TRANSPORT AND SIZE-SEPARATION OF AIRBORNE PARTICLES IN A MICROCHANNEL FOR CONTINUOUS PARTICLE MONITORING

by

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ABSTRACT

Microfluidics research and development has emerged as a novel and promising tool for the development of sensors and actuators. However, one area in which microfluidics has been only minimally employed is in the handling of airborne particles, or aerosols. The real-time monitoring of aerosols is important for protecting human health and earth’s environment. The small size of microchannels, coupled with the opportunity to integrate sensing technologies, suggests them as a promising tool for the next generation of aerosol sensors. To that end, this thesis presents a microfluidics-based system for the size-separation of aerosols. Specifically, centrifugal force is exerted on each particle as it travels around a curved microchannel, resulting in the particle occupying a size-dependent lateral position in the channel.

The behaviours of aerosols in a microchannel are examined, including the effects of flow focusing, the diffusion of airborne particles in a channel, and the centrifugal and viscous forces exerted on particles in a curved microchannel. Mathematical descriptions and computer simulations of these effects are developed to model these effects. Straight and curved microchannels were fabricated and each of these effects was measured experimentally, and compared to the models. Various combinations of airborne particles between 0.2 μm and 3.2 μm were successfully separated by size. A prototype optical particle detector was built and tested for its suitability as a candidate for integration with the microchannel particle separator.

This represents the first approach in which aerosols have been separated by centrifugal forces in a microchannel, and one of very few approaches that have been used for any kind of size-based separation of airborne particles in microchannels. The small footprint and potential for integration offered by microsystem fabrication technology make it a desirable avenue of pursuit for the development of small, portable particulate monitors. The results presented here confirm that this approach to size-separation is a feasible option for a future microsystem-based size-selective particulate monitor.
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LIST OF SYMBOLS

Particle physical properties
m  particle mass
ρ  particle density
d  particle diameter
D  radial distance traveled by the particle

Fluid physical properties
V  axial velocity
μ  fluid viscosity
T_{air}  fluid temperature
d_{air}  average diameter of an air molecule
λ  mean free path of air
ρ_f  fluid density

Channel geometry
r  bend radius
θ  bend angle
L  length (of channel)
W  channel width
H  channel height
D_H  hydraulic diameter
x, y, z  (length, height, width)

Experimental or calculated values
U  radial velocity
R  flow resistance
F  forces (various: F_{viscous}, F_{centrifugal}, etc)
P  pressure
Q  volumetric flow rate
S  ratio of sample width to total channel width (before diffusion)
Re  Reynold’s number
Kn  Knudsen number
De  Dean number
$V_{\text{Dean}}$  Dean flow velocity
p  probability distribution function for particle diffusion

**Particle generation and detection**

N  number of particles per second
M  change in mass of sample liquid
C  concentration of sample liquid, in w/v
$F_{\text{corr}}$  particle size dependent correction factor for nebulizer
A  area of filter exposed to particles
T  experiment run time
$R_{\text{SN}}$  signal to noise ratio
n  magnitude of the noise in the particle detector signal

**Other**

$k_B$  Boltzmann’s constant

a  particle detector photodiode output
1. INTRODUCTION

Microfluidics research and development has emerged as a novel and promising tool for the development of applications as varied as cell sorting [1]-[2], chemical and biochemical analysis [3], drug discovery [4], chemical reactors [5], and others [6]-[7]. These applications have been coupled with the parallel development of appropriate liquid handling tools such as micropumps [8] and valves [9] as well as the study of the fluid mechanics of such devices [10]-[11]. However, one area in which microfluidics has been only minimally employed is in the handling of airborne particles, or aerosols. The small size of microchannels coupled with the opportunity to integrate sensing technologies suggests them as a promising tool for the next generation of aerosol sensors. To that end, this thesis investigates the transport of aerosols in microchannels with a focus on modeling and measuring the effects of aerodynamic focusing, particle diffusion, and the centrifugal forces acting on particles in a curved microchannel. The final goal is to actuate a size-dependent movement of the particles in the channel to permit the separation of particles by size.

1.1 Aerosol properties and effects

The term “aerosol” has been defined as “an assembly of liquid or solid particles suspended in a gaseous medium long enough to be observed or measured” [12]. Aerosols span an extremely wide spectrum of properties, including physical size, shape, electrical properties, bulk materials, and surface chemistry. Some particles may consist of or be coated with materials that are particularly toxic, and these are frequently and necessarily the target of specific toxicity research, measurement techniques, and regulatory exposure limits which focus on one or more relevant properties. Furthermore, a material that is not toxic at the macroscale can also have negative health effects when it is present as a microscale particulate. For the purposes of classifying and regulating such so-called “nuisance dust”, the most common single parameter used is the particle’s size [12]. However, even this single parameter is subject to numerous approximations and corrections, as most common aerosols are not uniform in shape. To allow comparison between aerosols of different shapes and materials, various equivalent diameters exist which allow the properties of a particle to be normalized to that of an ideal sphere which would behave in the same way as the test aerosol under certain conditions. A common such equivalent diameter is the aerodynamic diameter of a particle, which is defined as the diameter of
a uniform 1000 kg/m$^3$ sphere which exhibits the same aerodynamic behaviour as the particle in question. Similarly, the equivalent optical diameter or equivalent electrical mobility diameter equate the test particle to a sphere that would scatter light or respond to an electrical field in the same manner, respectively [12].

1.1.1 Scaling effects

The forces which act on a particle can be compared to one another by their dependence on the length scales of the particle. For example, gravitational and inertial forces depend on the mass of a body, and therefore on the cube of the length scale. The viscous drag force on a particle moving through fluid scales with the length. As the size of the particle gets smaller, the forces with a higher dependence on length become less relatively significant to the behaviour of the particle.

Thus, the choice of size as a metric for aerosol monitoring and regulating is significant when examining health effects and outcomes of exposure. An aerosol’s size determines the most likely mechanism for its deposition within the respiratory tract and therefore how deep it will penetrate into the human respiratory tract and what tissues or physiological systems it will interact with [13]. The tracheobronchial region contains the airways that conduct inhaled air and matter from the nasal and oral passages to the alveolar regions where gas exchange takes place. Particles larger than 2-3 μm in equivalent diameter are highly likely to deposit by impaction in the tracheobronchial region, particularly in locations where there is an abrupt change in direction in the airflow. Smaller particles are more likely to avoid impaction by following the streamlines through to the alveolar regions. Here, particles can deposit by sedimentation (the effects of gravity) primarily if they are over 0.5 μm in diameter. Particles less than 0.5 μm in diameter are most efficiently deposited by diffusion, which can occur in all regions of the lungs and air passages [14]-[15].

1.1.2 Health effects and regulation of aerosols

Exposure to aerosols through inhalation has been shown to negatively affect human health, and has been linked to increased rates of asthma, cardiovascular disease, lung cancer, pulmonary diseases, and mortality [16]-[18]. The World Health Organization (WHO) estimates that exposure to outdoor air pollution is responsible for 1.4% of global mortalities and 2% of cardiopulmonary disease [19]. As a result of these negative health effects, occupational and health regulatory agencies worldwide have set limits on the recommended levels of environmental aerosols as well as occupational exposure limits. A common metric for both is
the PM2.5 level, or the concentration in air of particles with a diameter of less than 2.5 μm, typically measured in mass of aerosol per volume of air. The US Environmental Protection Agency and the WHO recommend that long-term PM2.5 levels not exceed 15 μg/m³ and 10 μg/m³ respectively, and the US Occupational Safety and Health Administration has a Permissable Exposure Limit of 5 mg/m³ for inert respirable dust over 8 hours [20]-[22].

1.2 Current techniques and technologies for measuring aerosols

Currently available commercial technologies used in environmental monitoring and occupational health applications use one of several principles [23]. Several of these techniques will be introduced here. Some of the earliest aerosol monitors were gravimetric samplers, in which particles from a large volume are collected on a filter that is weighed before and after the measurement. This technique was improved with the use of impactors and cyclones to select for particle size. Other approaches include optical sensors, which rely on the scattering of laser light to size and count individual particles or measure the concentration of a cloud of aerosol. Lastly, electrostatic separators and detectors cause particles to become electrically charged, and then select for particle size by manipulating the balance of electrostatic and inertial effects.

1.2.1 Inertial separators and collectors

Gravimetric techniques at their most basic offer particle sampling that is simple in concept and equipment. A pump pulls air through a filter at a known flow rate; careful measurements of the mass of the filter before and after the test provide an average mass of aerosol per volume of air sampled. This approach alone does not provide real-time data nor information on the size of the aerosol, but size data can be obtained through the inclusion of a size-selective device. Inertial-based size selection can be realized with the use of an impactor, which consists of stages of obstacles placed in the way of aerosol-laden fluid; particles smaller than a cutoff size are able to follow the fluid streamlines around the obstacle, while particles larger than the cutoff size are carried by their inertia onto the obstacle [24]. Whether a given particle will impact onto an obstacle is characterized by the Stokes number, which is a ratio of the stopping distance of the particle to a characteristic dimension of the obstacle. By controlling the size of various obstacle stages and the velocity of the air, several cutoff sizes can be achieved which allows for the distinction of several size bins of particles. A single cutoff size can be used as a pre-selector for a more complex particle analysis system, to remove particles larger than the size range of interest. If multiple cutoffs are used and the particles are collected for analysis, a major disadvantage of such a collection system presents itself: the obstacle itself must then be
weighed to determine the amount of matter that was deposited on it. Cyclones and virtual impactors both operate on a similar principle of selection by inertia, but rather than having an obstacle in the way of the flow, a sharp change in direction is imposed on the fluid by the geometry of the device. Larger particles are carried by their inertia across the sharply reversing streamlines, and follow a low flow path into a collection outlet. The system geometries and flow rates define the cutoff sizes of the separation process. Virtual impactors most significantly differ from ordinary impactors in that the lack of an impaction plate means that the particles larger than the cutoff size are merely redirected rather than impacted, allowing more flexibility in further analysis. However, a virtual impactor does not remove all of the small particles from the outlet flow, but simply increases the relative concentration of larger particles [25].

Lastly, the tamper element oscillating microbalance or TEOM (Thermo Scientific Model 1405) can measure aerosol mass directly and in real-time. A more advanced version of the gravitational technique, it collects particulate matter on a filter and measures the change in mass of the filter by measuring the change in its resonance frequency. This approach to counting is combined with a size-selective device to provide size-specific concentration measurements. It features high resolution and accuracy, but is sensitive to the temperature and humidity of the air [26]-[27]. Carefully controlling the sampling environment can lead to results that very closely correlate with those measured by a traditional impactor system [28]. At 18 kg and 1.5x10^5 cm^3, though, the TEOM is not suitable for portable sensing or personal exposure monitoring.

### 1.2.2 Optical size detectors and counters

Optical particle counters rely on a measurement of the change in intensity of scattered or absorbed light to detect and measure the size of particles. Such devices can provide continuous real-time data, but only provide an optical (rather than aerodynamic) equivalent diameter, and are sensitive to the aerosol’s optical properties.

One approach to optical detection has a beam of laser light focused on a region through which particle are passed. As each particle passes across the beam, light scattered from the laser is measured with photodetectors placed around the light’s path. The amount of scattered light depends on the particle’s size, as well as its density, shape, and refractive index [29]. A limitation of this approach lies in the change in scattering behaviour of particles that are significantly smaller than the wavelength of the light. This limitation can be overcome with the use of a condensation nucleus growth chamber, in which the particle’s apparent size in increased by the condensation of a saturated vapour onto the particle. Further, the possibility of multiple
particles simultaneously passing through the measurement region adds uncertainty to the size measurement.

A second approach to optical detection is to forgo individual particle measurements in favour of an overall measurement of the density of a cloud of particles by the measurement of light scattered from a volume of air. This mass-measurement approach, known as nephelometry, allows for the measurement of a wide range of particle concentrations, but lacks the high size-resolution of the single particle counter [29].

Another variation on the optical detection theme is that of the aethelometer [30]-[31] or the particle soot absorption photometer (PSAP) [32]. Both devices run particle-laden air through a filter and measure the change in optical transmission of the filter as it accumulates particles. This approach is primarily used for measuring carbon black soot, which strongly absorbs light. The change in intensity as the soot is collected is related to the absorption of the particles, providing information that can be correlated back to an estimated mass concentration.

The Dusttrak DRX Aerosol Monitor (TSI Inc, Model 8533 and 8534) combines the single particle measurement approach with nephelometry in a single instrument to take advantage of the benefits of each [33]. All optical instruments, however, are subject to the limitations of the aerosol’s usually unknown optical properties. The device must be calibrated for each type of aerosol if a correlation between optical and aerodynamic equivalent diameter is to be made accurately.

### 1.2.3 Electrical separators and counters

Electrical approaches to particle separation select particles based on the ratio of applied electrical and mechanical forces on a particle [34]. Particles are given an electrical charge and then transported through an electrical field; through this process, particles are sorted according to their electrical mobility, defined as a particle’s velocity relative to the fluid medium divided by the electrical field strength. Smaller particles, for which the electrical attraction quickly overcomes inertia, will be attracted to the system’s electrodes, while larger particles with more inertia will pass through to an outlet. A particle’s electrical mobility depends on the properties of the fluid medium, the particle size, and the number of charges on the particle [35]. Electrical separation systems are particularly suited to sub-micron particles, as the size of the electric field required for sorting increases with particle size. However, the rate of diffusion increases with decreasing particle size, causing a worsening of the achievable resolution of the separation. Electrical systems are often more tunable than purely mechanical systems, as the sorting
characteristics are not only a function of the mechanical geometry but also of the more easily adjusted electrical parameters.

One of the most difficult aspects to electrical separators is that of charging the test particles [35]. Particles must first be neutralized, and then charged through interactions with ions in a gaseous medium. The production of these ions is usually achieved through a high-voltage corona discharge or through radioactive decay, and photoelectron emission and electrospray droplet formation methods have also been used. For an ideal system, each particle would have the same total electrical charge to permit sorting only by particle size. However, every charging system will result in some variation in the number of charges per particle. Work has been done on predicting and measuring the distribution of charge on particles or agglomerates for various types of charging systems, so that the apparent distribution resulting from an instrument’s measurements can be probabilistically corrected to reflect the actual particle sizes present in the sample [36]-[38].

1.3 Microfluidics for particle separation

Microchannels have been used for the separation by size or other property of microscaled particles, but the majority of this work has focused only liquid-borne particles. Primarily aimed at biological applications, such work has used techniques relying on flow-field fractionation, inertial lift forces, dielectrophoretic forces, electrostatic forces, optical sizing, and others methods. Some mechanical approaches to liquid-borne particles separation in microchannels will be discussed briefly, as they are the most analogous to the airborne separation approach used in this project. Airborne particles have also been sorted or counted in microchannel using electrical and mass measurement techniques, and a review of these approaches will also be presented below.

1.3.1 Particle separation in liquid in microchannels

Flow field fractionation techniques are somewhat analogous to chromatography, as they separate particles by their rates of transport through a system. A size- or other property-dependent rate of transport is realized by the application of an externally applied field, which could include gravitational, magnetic, electrical, or thermal forces. This force causes some particles to move out of fast-moving regions of the flow, thereby retarding their progress through the system. This approach often has a limited flow rate, since the time required for the external field to act upon the particles to be sorted limits the speed of bulk transport through the device [39]-[40].
The focusing of particles in liquid microchannel has also been accomplished using inertial lift forces, which cause particles to occupy a lateral equilibrium position in the channel [41]-[42]. These equilibrium positions are not dependent on the particle size [43]. However, the axial distance required for a particle to reach its equilibrium position does depend on the particle size, and this effect was used to achieve separation of 3 μm and 10.2 μm particles in a microchannel, although separation efficiencies were less than 70% [44]. The effects of these inertial lift forces have also been combined with the size-dependent drag caused by secondary Dean flows, causing particles to occupy a size-dependent position in the cross-section of spiral or curved microchannels. The spiral channel showed 90% separation efficiency for a mixture containing 10 μm, 15 μm, and 20 μm diameter particles [45] and complete separation of a mixture of 1.9 μm and 7.32 μm particles [46]. The curved channel, which incorporated several repeating asymmetrical serpentine shapes, was tested with a mixture of 3.1 μm and 9 μm particles; after two cascading separations, 99.9% of the 9 μm particles had been separated out of the mixture but only 56% of the 3.1 μm particles were still present [47]-[48]. Also demonstrated was the separation of 4.5 μm particles from 3 μm particles with 90% efficiency, but again half of the smaller particles were present in each final sample.

Particles have also been sorted in a microchannel by dielectrophoresis, which causes the motion of electrically neutral matter in a nonuniform electric field due to polarization effects. This approach has been used by several groups, particularly for on-chip cell sorting [1]. One study has demonstrated the separation of airborne bacteria from polystyrene latex microspheres by electrophoresis in a microchannel; however, this required the introduction of the airborne bacteria into a liquid before separation [49].

### 1.3.2 Airborne particle separation or detection in microchannels

Airborne particles in a microchannel were sorted by electrical mobility in [50]. Particles were charged with a microfabricated corona ionizer using a high-voltage pin between planes. The microfabricated device required high voltages (>1kV) for charging and later collecting the particles, which would limit portability, and the fabrication was a complex process requiring chemical etching. The device was designed to selectively allow particles to pass through the electric field of the collection region, with the larger particles (i.e., those with lower electrical mobility) passing through the region while the smaller particles with higher electrical mobility were trapped by the field. The device was tested for particles ranging in size from 50 to 122 nm, and the proportion of particles passing through the collection region ranged from 0.225 to 0.32
for the smallest to largest particles respectively. While these results proved the general concept of size-selective collection, the selectivity is too low to be of practical use within the range of sizes presented in the work. It is reasonable to expect, however, that the device could usefully remove particles below a certain size if a wider range of particle sizes had been tested.

Airborne particles have also been selectively concentrated by size in a micro virtual impactor [51]. A simple channel consisting of an inlet and three outlets was fabricated, with all of the channels at 90° to each other. The two outlets that were at 90° to the inlet (the major flow outlets) together had 90% of the flow volume exit the channel through them. The third inlet (the major flow outlet) had the remaining 10% of the flow exit through it. Particles larger than a geometrically-defined cutoff would be carried forward by their inertia past flow diverting into the major flow outlets and would exit out the minor flow outlet; particles smaller than the cutoff would be present in both the minor and major flow outlets. This led to a selective increase in concentration of particles larger than the cutoff size in the minor outlet flow. The primary disadvantages of this system are that the minor flow, while containing nearly all of the particles larger than the cutoff, also contained around 10% of the particles smaller than the cutoff size. Additionally, high rates of particle loss (>30%) onto the walls were reported. The cutoff size of the system is a function of the system geometry and particles, but is also inversely proportional to the square root of the average velocity at the end of the inlet. To change the system to select for a different size of particle would require either a change in geometry or a change in velocity inversely proportional to the square of the change in cutoff diameter.

Black et al. [52] presented a mass detection based sensor for measuring the concentration of particulate matter. Thin-film bulk acoustic wave resonators were fabricated and the mass of deposited particulate matter was measured by monitoring the resulting change in the resonant frequency of the device. The particulate matter was deposited on the surface of the resonators by thermophoresis, which causes a net motion of particles away from a heat source due to the unequal amounts of thermal energy on the near and far side of the particle from the source. This detection system did not have any size-selective behaviour but only measured a total change in mass. The system was tested in a controlled environmental chamber with tobacco smoke and provided data that was well-correlated with purchased particulate matter monitors. This approach provides high resolution and sensitivity to concentration levels – it showed a limit of detection of 18 μg/m³ and was sensitive to mass changes down to 1 pg – but it would be most useful if combined with a continuously operating size-selective device to characterize not only the concentration but also the size of the test aerosol.
In general, one of the most sought-after advantages of microfluidics is the potential to take advantage of the low-cost bulk fabrication processes that have enabled the meteoric development of microelectronics [6]. Such fabrication processes are generally characterized by the small size of achievable device features and high level of on-chip integration, but often at the cost of ease of interconnection with the “macro-scaled” world. Thus, a microfluidics chip approach – one that does not require high power or sample preparation, and one that does take advantage of integrating particle separation, detection, and data storage or presentation – is a potentially potent tool for monitoring personal exposure to aerosols or high-resolution spatial monitoring of particulate matter levels.

1.3.3 Our approach to measuring aerosols

This project develops a microfluidics-based inertial system for the size-separation of aerosols. Specifically, centrifugal force is exerted on each particle as it travels around a curved microchannel, resulting in the particle occupying a size-dependent lateral position in the channel. The description of the system, including the fluid mechanics governing the flow and the particle’s behavior in it, are developed in Chapter 2, along with a description of the principles of detection. Simulations modeling the behaviour of the particles in the channel are presented in Chapter 3. The separation device is fabricated with the standard photolithography techniques used in microfluidics, as described in Chapter 4. The device is tested with polystyrene latex (PSL) microsphere; Chapter 5 outlines the experimental setup and processes in general, including the properties and generation of the test particles and how they are interfaced with the microchannels. Experimental results are presented and discussed in Chapter 6. Chapter 7 presents the results of simulations which were undertaken to provide a framework for optimization of future designs and operating parameters, and the overall conclusions and further work are in Chapter 8.
2. DEVICE PRINCIPLES AND DESIGN

Airborne particles entering a microchannel are focused towards the channel centre by two sheath air inlets before travelling around a curved section of the channel where competing centrifugal and viscous forces exert a net radially outward size-dependent movement on the particles (Figure 1). The particles are trapped on a filter placed in the outlet so their position in the channel can be recorded and measured. Two designs were fabricated; one with a single outlet, for obtaining a measurement of particle positions continuously across the channel width, and a second design with the outlet splitting into two or three sub-channels (as in Figure 1) causing the particle-laden air to split into two or three separate streams, with each stream containing a different size range of particles.

Figure 1. Principle of operation of curved separation channel

Three effects in the main channel section are studied: the width of the sample stream in the main channel section, the diffusion of the particles out of the sample stream and into the sheath streams, and the radially outwards motion of the particles due to the centrifugal force. Each affects the separation ability of the system and informs the design decisions, therefore, mathematical models for each of these attributes are presented in this chapter, with simulations of each presented in Chapter 3. The experimental setup and results demonstrating each are in Chapters 5 and 6.
2.1 Flow focusing to control the width of the sample stream

Particles are confined to a region in the centre of the channel through flow focusing resulting from the air in the sheath flow inlets (Figure 1). To determine the width of the resulting sample stream containing particles, the flow profile in the main channel section and the flow resistances of the channel sections must be known.

The characteristically small dimensions of microchannels typically lead to flows with low Reynolds numbers

\[ \text{Re} = \frac{\rho_f V D_H}{\mu} \]  

(1)

depending on the fluid density \( \rho_f \), viscosity \( \mu \), and velocity \( V \); the hydraulic diameter \( D_H \) of a rectangular channel of width \( W \) and height \( H \) is given by \( D_H = 2WH/(W+H) \). In the case of low Reynolds numbers, the inertial terms of the Navier-Stokes equations can be neglected, leading to the equation of flow

\[ \frac{\partial^2 u}{\partial y^2} + \frac{\partial^2 u}{\partial z} = \frac{1}{\mu} \frac{\partial P}{\partial x} \]  

(2)

for the axial velocity \( u \), which is a function of the position \( (y, z) \) in the channel cross section and of the pressure \( P \) as a function of the axial position \( x \). Since the cross-sectional shape of each section of the microchannel system is uniform, the pressure drop is taken to be linear with axial position, and the pressure gradient \( \frac{\partial P}{\partial x} \) becomes simply \( -\frac{P_0}{L} \), the total pressure drop \( P_0 \) divided by the channel length \( L \). To solve equation 2 for \(-H/2<z<H/2\) and \(-W/2<y<W/2\), the no-slip condition is imposed on the walls as the boundary condition \( U(\pm W/2, z) = U(y, \pm H/2) = 0 \), yielding the flow profile \([53]\)

\[ U(y, z) = \frac{4H^2 P_0}{\mu \pi^3 L} \sum_{n \text{odd}}^{\infty} (-1)^{n+1} \frac{n-1}{2} \frac{\cos \left[ \frac{n\pi y}{2H} \right]}{n^3} \left\{ 1 - \frac{\cosh \left[ \frac{n\pi y}{2H} \right]}{\cosh \left[ \frac{n\pi W}{2H} \right]} \right\} \]

(3)

shown in Figure 2.
The flow resistance $R$ is defined as the total pressure drop $P_0$ along the channel divided by the flow rate $Q$ through the channel; it is analogous to electrical resistance, which is the voltage drop divided by the current. For a rectangular channel, the resistance per length

$$\frac{R}{L} = \frac{P_0}{QL} = \frac{12\mu}{H^3W} \times \left[1 - \frac{192H}{\pi^5W} \sum_{n_{odd}}^\infty \left\{\frac{\tanh\left(\frac{nnW}{2H}\right)}{n^5}\right\}\right]^{-1}$$

is found by integrating the velocity profile (equation 3) to find the total flow rate, dividing by the pressure drop, and inverting the equation [53]; the results are shown in Figure 3, with air as the fluid with $\mu=1.8\times10^{-5}$ and a channel height of 100 $\mu$m.
The sheath and sample air inlets will all be pressure-controlled; the flow resistances of the individual channel sections are used to calculate the flow rates $Q_i$ of the sheath and sample air as a function of the driving pressures $P_i$ with a fluidic version of Ohm’s Law (Figure 4) [54]. For four channels which connect at one node, the total mass flow into the node must be equal to the mass flow out

$$Q_1 + Q_2 + Q_3 + Q_4 = 0$$

and since

$$Q_i = \frac{P_i - P_c}{R_i}$$

with $P_c$ the pressure at the node where the inlets join together (Figure 4), the inlet pressures and channel flow resistances can be related as

$$\frac{(P_1 - P_c)}{R_1} + \frac{(P_2 - P_c)}{R_2} + \frac{(P_3 - P_c)}{R_3} + \frac{(P_c - P_{atm})}{R_4} = 0$$

with $R_1$ the flow resistance of the sample channel, $R_2$ and $R_3$ that of the sheath flow channels, and $R_4$ that of the main outlet channel plus any other fluidic resistances; for example, that of a filter collecting particles at the channel exit.
Equation 7 yields an expression for the pressure at the node

\[ P_C = \left( \sum_{i=1}^{4} \frac{P_i}{R_i} \right) \times \left( \frac{R_1 R_2 R_3 R_4}{R_1 R_2 R_3 + R_1 R_2 R_4 + R_1 R_3 R_4 + R_2 R_3 R_4} \right). \]  

(8)

This pressure \( P_C \) is used in equation 3 to obtain the flow profile \( u(y,z) \) in the main channel (channel section 4, in Figure 4). Knowing \( P_C \) provides the flow rates \( Q_i \) in each section of the microchannel system (equation 7). In steady-state conditions, the flow rate in the sample air section and in each sheath section of the main section of the channel must equal the flow rate of the inlet providing air to that section of the flow. Since the flow velocity is not uniform in the cross-section of the main channel section, the width of the sample stream must be calculated using the known flow rates of each inlet and the velocity profile in the main channel. The width of each sheath air section \( W_{\text{sheath}} \) is found by equating the flow rate of one sheath air section in the main channel to the flow rate of one sheath air inlet

\[ \frac{\int_0^{W_{\text{sheath}}} \int_0^h u(y,z) \, dy \, dz}{\int_0^W \int_0^h u(y,z) \, dy \, dz} = \frac{Q_S}{2Q_S + Q_1} \]  

(9)

for \( Q_S = Q_2 = Q_3 \). Knowing the total width of the channel \( W \) and the width \( W_{\text{sheath}} \) of each of the two sheath air streams gives the width of the sample stream

\[ W_{\text{sample}} = W - 2 W_{\text{sheath}}. \]  

(10)
2.2 Diffusion of aerosols

Gas molecules are in constant motion because of their own thermal energy. As they impact a larger molecule or particle, it undergoes random motion larger than the motion from its own thermal energy. This random motion resulting from impaction by gas molecules is known as Brownian motion. As an ensemble of concentrated molecules or particles experiences Brownian motion, the net result is a time-dependent stochastic diffusion process by which the region of concentrated particles will gradually spread out. This phenomenon is relevant to the present application, as this diffusion resulting from Brownian motion will cause the stream of sample particles to gradually spread into the sheath air. This spread is described by the solution to the Fokker-Planck equation

$$\frac{\partial p}{\partial t} = -\nabla \cdot (\gamma p) + \nabla \cdot (D \nabla p),$$

which describes the evolution in time of a probability density function $p$ of the position of a particle about a mean $\gamma$ and translational diffusion coefficient

$$D = \frac{k_B T C_S}{3 \pi \mu d}.$$  

The diffusion coefficient of a particle depends on the particle and fluid characteristics as well as on the Boltzmann constant $k_B$, the temperature $T$ in Kelvin, and the Cunningham slip correction factor

$$C_S = 1 + \frac{2 \lambda}{d} \left( A_1 + A_2 \exp \left( \frac{A_3 d}{\lambda} \right) \right),$$

which is a correction to the drag force predicted by the Stokes equations. The derivation of the factor was derived for spheres by E. Cunningham in in 1910 and it is used when particles are sufficiently small, or the fluid pressure sufficiently low, that the medium in which particles travel may no longer appear fully continuous – that is, when the mean free path $\lambda$ of the fluid molecules is comparable to the particle size [55]. For spherical particles traveling in room temperature air, the Cunningham slip correction factor is $C_S = 1.52$ for 0.5 $\mu$m diameter particles and $C_S = 1.085$ for 3.0 $\mu$m diameter particles. Since the diffusion coefficient is linearly dependent on the correction factor, the factor directly affects the expected rate of diffusion of particles of a given size.

Since there is no preferred direction to the diffusion process, the Fokker-Planck equation is solved for $\gamma=0$. In one dimension, the resulting probability
\[ p(y, t, 0, 0) = \frac{1}{\sqrt{4\pi D t}} \exp\left(-\frac{y^2}{4Dt}\right) \] (14)

that a particle located at position \( y = 0 \) at time \( t = 0 \) will be found at position \( y \) at time \( t \) is a normal distribution with a mean of zero and a standard deviation \( \sigma = \sqrt{2D t} \) [56]. Since the mean position of a large ensemble of particles will be zero at all times, the root-mean-squared (RMS) distance travelled by the average particle \( < L > = \sqrt{\sigma^2} = \sqrt{2D t} \) is used to represent the amount of diffusion experienced by an ensemble of particles. In two dimensions, the probability distribution extends to

\[ p(y, z, t, 0, 0, 0) = \frac{1}{\sqrt{16\pi^2 D t}} \exp\left[-\frac{(y^2 + z^2)}{8Dt}\right] \] (15)

with the resulting RMS distance \( < L_{2D} > = \sqrt{4D t} \) . Particles beginning at one point on a plane will disperse outwards with a two-dimensional Gaussian probability distribution (Figure 5).

![Figure 5. Graphical representation of the two-dimensional probability distribution function resulting from the diffusion of 1 μm particles in air from a point source at the centre of the image after a time of (a) 0.1 s, (b) 1 s, (c) 10 s. Each image is 100 μm by 100 μm.](image)

This model is applied in section 3.2 to simulate the spreading of the sample particles into the sheath air due to diffusion.

2.3 Displacement of aerosols by centrifugal forces

To realize a size-specific motion of the particles in the sample air, a curved microchannel was used to impart centrifugal forces on the test aerosols (Figure 1). For a sphere of mass \( m \), density \( \rho \), and diameter \( d \) moving in an arc of radius \( r \) with velocity \( V \), the centrifugal force
The particle reaches a velocity that is within 1% of the fluid’s velocity within 153 μs (Figure 6).
The distance travelled by the particle while it is accelerating to the speed of the fluid is found by integrating the velocity (Figure 7); for the 3.2 μm particle data shown here, the particle travels less than 1 mm axially before it reaches 99% the velocity of the fluid.

This estimate is conservative in a number of ways. 3.2 μm is the largest particle size used in the experiments, and is therefore the slowest of the available particles to reach full speed. The
particles are not starting from rest, since they are moving in the sample inlet before reaching the main channel section, so the test particles would need even less distance than 1 mm to reach full speed. The design used in this project has a 2 mm straight stretch after the joining of the inlets and before the curve.

The balance of the centrifugal and viscous drag forces

\[ F_{\text{centrifugal}} = F_{\text{viscous}} \]

allows the calculation of the radial velocity

\[ U(y, z) = \frac{1}{18} \left( \frac{\rho d^2 V(y, z)^2}{r \mu} \right) \]  
(19)

of the particle at any point in its movement around a bend, as long as the fluid velocity \( V \) is known at that point (Figure 8).

![Graph showing air axial velocity and particle instantaneous radial velocity](image)

**Figure 8.** Flow profile of air at the mid-height point of a 60 μm wide microchannel, and the corresponding radially outward velocity of a 1 μm particle experiencing the centrifugal forces imposed by the flow in channel with \( r=2.5 \) mm, \( \mu=1.8\times10^{-5}, \rho=1050 \) kg/m³.

The variance of the radius of curvature caused by the finite channel width was neglected and instead the radius of curvature of the centre of the channel was used. The maximum error in
radius of curvature due to this simplification is just over 1%. The radius of curvature, viscosity, and density of the particles are all assumed to be constant. As seen by equation 19, the radially outwards velocity of the particle is a function of the particle diameter squared, which corresponds to the more qualitative scaling of forces described in section 1.1.1.

2.4 Other effects on the flow

2.4.1 Dean flow

In curved channels or pipes, it is possible for a secondary flow known as the Dean flow to occur. The first order Dean flow can be seen in the form of a cross-sectional flow travelling radially outwards in the channel centre, with the flow travelling back towards the radially inwards side along the top and bottom of the channel. The magnitude of the flow is determined by the dimensionless Dean number

\[ \text{De} = \frac{\rho V D_H}{\mu} \sqrt{\frac{W}{2r}} = \text{Re} \sqrt{\frac{W}{2r}} \]  

(20)

The Dean flow has been used to actuate size-specific movements of particles in microchannels with Dean numbers of between 1 and 10 \[47\][57]. In the range of 10<De<40, the average velocity of the Dean flows in the channel cross-section was found to be \(V_{\text{Dean}} = 1.8 \times 10^{-4} \text{De}^{1.63} \text{ m/s} \) [58]. In all experiments performed in this work, the Dean number lies between 0.1 and 0.8, and the Dean flow was not included in the simulations. If this average Dean velocity is applied to obtain an estimate of the possible error caused by neglecting the Dean flow, the resulting radial displacement due to Dean flows is 0.44±0.25 μm in the range of flow conditions used in this work.

2.4.2 Slip flow

The validity of the no-slip boundary condition which was used to find the flow profile in the channel is determined by the dimensionless Knudsen number, defined as the ratio of the mean free path of fluid molecules to a characteristic physical dimension. The mean free path of air

\[ \lambda = \frac{k_B T_{\text{air}}}{\sqrt{2 \pi d_{\text{air}}^2 P}} \]  

(21)
depends on its temperature $T_{\text{air}}$, pressure $P$, and the average diameter of a molecule of air $d_{\text{air}}$. Several regimes of flow exist depending on the Knudsen number; a set of commonly cited values and their respective models are shown in Table 1.

<table>
<thead>
<tr>
<th>Knudsen number range</th>
<th>Model typically used for analysis</th>
</tr>
</thead>
<tbody>
<tr>
<td>$&lt; 0.01$</td>
<td>Navier-Stokes equations with no slip</td>
</tr>
<tr>
<td>$0.01 &lt; \text{Kn} &lt; 0.1$</td>
<td>Navier-Stokes equations with boundary slip flow</td>
</tr>
<tr>
<td>$0.1 &lt; \text{Kn} &lt; 10$</td>
<td>transition regime; various methods exist</td>
</tr>
<tr>
<td>$\text{Kn} &gt; 10$</td>
<td>rarefied flow; must be analyzed with statistical methods</td>
</tr>
</tbody>
</table>

Some literature suggests that slip flow conditions may start having discernable effects on the flow as low as $\text{Kn} = 0.001$ [10][60]. For a 60 µm by 100 µm rectangular channel with air at standard temperature and pressure,

$$\text{Kn} = \frac{\lambda}{D_h} \approx 9 \times 10^{-4}$$  \hspace{1cm} (22)

This low Knudsen number validates the use of the no-slip condition as a boundary condition of the velocity profile.

### 2.5 Detection principle

A prototype particle detector which uses the same principle as the aetheolometer and PSAP methods introduced in section 1.1.2 was built to test for its potential for future integration with the particle sorter. Similarly to these commercially available technologies, our detector is a filter-based particle sensing device which captures particles on a filter, with the operation principle demonstrated in Figure 9. A light-emitting diode (LED) is placed upstream of the filter and shines light through the filter. A photodiode downstream of the filter emits a voltage in response to the intensity of the light passing through the filter. As particles are trapped on the filter, the light incident on the photodiode’s surface becomes less intense. The output of the
photodiode is recorded, and the change in intensity is related to the time-integrated total particle concentration [61]

Figure 9. Principle of operation of the particle detector
3. SIMULATIONS

3.1 Simulations of the flow focusing of the sample stream

MATLAB (v. 7.8.0 R2009a, The MathWorks Inc.) was used to solve Equations 3, 4, and 9 to find the width of each section of sheath air. This value was used to find the ratio of sample to sheath air flow rates (Figure 10) and then the corresponding width of the sample stream in the main channel for any given set of inlet pressures.

![Graph showing width of sample stream vs. ratio of sheath air to sample air flow rates](image)

**Figure 10.** Width of sample air as a function of ratio of sheath air to sample air by flow rates

3.2 Simulations of particle diffusion

There are multiple ways to model the effects of diffusion on individual particles. A Langevin equation

\[
\frac{\Delta U}{\Delta t} = F_{\text{viscous}} + F_{\text{Brownian}} + F_{\text{others}}
\]  

(23)

includes the Brownian motion modeled as a stochastic force

\[
F_{\text{Brownian}} = G_{i} m \frac{216 \mu k_{B} T}{\Delta t \rho^{2} S^{2} d^{5} C_{c}}
\]  

(24)

with \( G_{i} \) a Gaussian normal distribution with zero mean and unity standard deviation, \( S \) the ratio of particle to fluid densities, and \( \Delta t \) a time increment over which the force is applied [62][63].
somewhat simpler approach treats the Brownian motion as a random walk superimposed on the force-based deterministic motion of the particle. This application of the double random walk to the model of particle behavior in a straight channel begins by placing a number of particles uniformly across the sample air position at the beginning of the channel. This uniform distribution of particles is determined by the width of the sample stream as calculated by equation 11; a simulated particle is placed every 0.3 μm in the height and width dimensions in the sample stream region. Each individual particle is followed along the channel in small time intervals Δt. At each time interval two random numbers

\[ \alpha = \text{rand}(0,2\pi) \]

\[ \beta = \text{normrnd}(0,\sqrt{4D\Delta t}) \]

are chosen, where \( \text{rand}(0,2\pi) \) represents a random number chosen between 0 and 2π, and \( \text{normrnd}(0,\sqrt{4D\Delta t}) \) represents a random number chosen with the a normal probability density function with mean 0 and standard deviation \( \sqrt{4D\Delta t} \). From these, the particle’s incremental displacement

\[ \Delta x = V(y, z) \Delta t \]

\[ \Delta z_{\text{Brownian}} = \beta \cos(\alpha) \]

\[ \Delta y_{\text{Brownian}} = \beta \sin(\alpha) \]  

is determined. Any particle that was brought by centrifugal forces or diffusion to a distance of less than its own radius to the wall was removed from the simulation. Once all of the particles had been simulated and their final positions recorded, a cross-sectional image was generated where the darkness of the image in a location corresponds to the number of particles ending their simulation at that location. Sample results from the simulation of particle behavior in a straight channel show the widening of a sample stream of particles into sheath air regions due to Brownian motion, with the effect more significant for smaller particles and at lower velocities (Figure 11). The widening of the sample stream is more pronounced at the top and bottom of the channel, where the low velocity near the wall means that each particle in that region has a longer residence time in the channel, and therefore a higher mean distance traveled by Brownian motion.
Figure 11. MATLAB simulation of aerosol diffusion in a straight microchannel, with a top view of the whole channel, a cross-sectional view of the outlet, and a plot of concentration of particles as a function of position across the channel. Mean velocities and particle diameters are as labeled on top. The channel width is 60 μm, height is 100 μm, and length is 4 cm; dimensions are not to scale in the images.
3.3 Simulation of combined effects of centrifugal force and diffusion

The simulation of particle behaviour in a curved separation channel was similar to that of diffusion alone in a straight channel, except that along with the effects of diffusion and axial transport, at each time interval the local fluid velocity determines the radial balance of forces on the particle. The resulting new particle position in the axial and radial directions was calculated before the random walk from Brownian motion was applied. The combination of the Brownian motion model with the balance of the forces in Equation 17 yields the change in position of the particle at each time interval

\[
\begin{align*}
\Delta x &= V(y, z) \Delta t \\
\Delta y &= \Delta y_{\text{Brownian}} \\
\Delta z &= \frac{1}{18} \left( \frac{\rho d^2 V(y, z)}{\mu} \right)^2 \Delta t + \Delta z_{\text{Brownian}}
\end{align*}
\]  

(27)

Throughout the simulation, any particle that had a distance to the wall of less than its own radius was removed from the simulation; further discussion on the effects of this boundary condition can be found in Appendix 1. As with the straight channels in the previous section, the final cross-sectional positions of all particles were recorded, and the values used to generate a cross-sectional image of the position of the particles in the channel (Figure 12 a & b, top). The number of particles was summed up along its columns to produce a two-dimensional graph of particle position as a function of the width in the channel (Figure 12 a & b, bottom.) Detailed sample results are shown for the separation of 0.75 μm and 1.9 μm particles (Figure 12), and a summary of the results of separation of 0.2 μm, 0.75 μm, 1 μm, and 1.9 μm particles is given in plot form (Figure 13). From the cross-sections shown in Figure 12 it can be seen that the low flow regions at the top and bottom of the channel are causing significantly lower lateral position shifts in those regions than in the height-wide centre of the channel. In the mid-height regions, the particles are distinctly separated for both of the velocity conditions shown, but the lower velocity case has significantly more overlap in the particles at the top and bottom, leading to a lower separation resolution than the higher velocity. It is also clear that as expected, the smaller particles have a higher rate of diffusion than the larger particles, causing the band of smaller particles to be more spread out than that of the larger ones.
Figure 12. MATLAB simulation results of 0.75 µm (light green) and 1.9 µm (dark blue) particles undergoing separation and diffusion with (a) $v=0.25$ m/s and (b) 0.5 m/s, both at a sheath to sample ratio by flow rate of 5:1. Each picture is a sectional view of the outlet with a plot of concentration of particles as a function of position across the channel for that distribution. The channel width is 60 µm and height is 100 µm. The particles travel around a curve (radius of curvature 2.5 mm, angle of curvature 180°) and then travel a straight path after the end of the curve; total channel length is 4 cm. Image dimensions are not to scale.
The separation summary (Figure 13) provides a more general overview of the separation characteristics of the microchannel for a wider range of particle sizes and average flow velocities. At the end of each simulation, the ensemble of all the mean particle position in the channel width was calculated. At the lowest velocity shown, the three smallest particles are grouped into a relatively narrow region near the middle of the channel while the largest particle size has moved on average more than half of the distance between its starting position at the centre of the channel to the radially outwards wall. As the mean flow velocity increases, the particles spread out more until their mean positions are more evenly distributed across the radially outer half of the channel. Increasing the velocity even further leads to the similar grouping near the outer wall as was seen near the centre of the channel at low velocities. With the velocity increases, the mean position of the largest particles stops moving outwards; this is due to these simulated particles reaching the outer wall of the channel, an occurrence explored further in the following section.

Figure 13. Results of separation simulation, showing the mean position of each particle size. Each simulation had a sheath to sample ratio by flow rates of 3:1.
3.3.1 Simulation of particle deposition on the channel walls due to diffusion and centrifugal forces

Airborne particles traveling in microchannels can deposit on the walls of the microchannel in one of four ways. Electrostatic attraction between the particles and the channel walls can cause deposition, particles can diffuse onto the walls, gravitational forces can cause the particles to settle, and excessive centrifugal forces created at the channel bend can cause the particle to impact against the radially outer wall. The latter three of these four were modeled and the results used to inform experimental parameters and channel properties whenever possible.

The rates of particle deposition on the channel walls due to diffusion and centrifugal forces were obtained from the simulations described above (Section 3.3). Specifically, as the simulations followed individual particles down the channel, the number of particles coming to within one particle radius of a wall were counted, with an initial ratio of sheath air to sample air by flow rate of 3, mean channel velocities of 0.5, 1, and 1.5 m/s, and particle diameters of 0.2 μm, 1 μm, and 1.9 μm. This simulation assumed that any particle coming within one particle radius of a wall would adhere to that wall. The results clearly show the impaction of 1.9 μm particles against the radially outer side wall due to centrifugal forces, with over 70% of the simulated particles deposited on the radially outer wall at velocities of 1 and 1.5 m/s. In addition, 4-7% of the 1.9 particles are deposited on the top and bottom walls. The 0.2 μm particles, with a larger diffusion rate, have a deposition rate of 9-14% on the top and bottom walls (Figure 14), but negligible side wall deposition due to the relatively low centrifugal forces on these particles. While the assumption of adhesion based on proximity represents a worst-case scenario, the significant jump in proximity adhesion events shown by the 1.9 μm particles between the lowest velocity and the higher two velocities is telling. Even if not all of the critical particles adhere, the ones that do will make the channel effectively narrower in the region where they are located, thus increasing the chances of later particles getting caught in the same location. These results, despite the assumptions, provide an upper limit to what range of velocities are feasible to use with a given particle size, and this guideline can be applied and confirmed in experimental work.
Figure 14. Percent of simulated particles that deposit on the channel walls during simulations of a curved channel. Particle size is as labeled; the ratio of sheath to sample air is 3; the channel is 60 μm wide and 100 μm high with a 2500 μm radius of curvature.
4. DEVICE FABRICATION

Microchannels for characterization and separation experiments were fabricated using the standard photolithography techniques common in microfluidics (Figure 15). The photolithography process produces a negative mold of the channel, which in Figure 15 has a height $H$ and width $W$. The mold is then used to form the channel out of an optically transparent silicone rubber, polydimethylsiloxane (PDMS).

4.1 Overview of the photolithography process

The photolithography process took place in the UBC Cleanroom’s Class 1000 yellow room. The Class 1000 designation means that the room has less than 1000 particles larger than 0.5 microns per cubic foot of air; the yellow light allows for the use of photoresist polymers which are sensitive to light of shorter wavelengths. The microchannel mold was made from one

![Figure 15. Fabrication process overview. Each image is a cross-section of the horizontal layers of wafer, photoresist, or PDMS.](image-url)
of two photoresist polymers, SU-8 2075 or SU-8 3050 (MicroChem Corp., Newton, MA, USA) patterned on a silicon wafer substrate. The recipe used to form the SU-8 pattern is based on the datasheet provided by the photoresist manufacturer [64]. After being cleaned and coated with a thin layer of photoresist (Figure 15-1), the wafer is baked, exposed to ultraviolet light through an optical mask, and baked again. This process causes the selective curing of the photoresist in the regions where the optical mask allowed the photoresist to be exposed to ultraviolet light (Figure 15-2). Specifically, the UV light causes a strong acid to be formed in the optically exposed areas of the photoresist, and the following post-exposure baking step induces an acid-catalyzed cross-linking of the photoresist polymer, driven by thermal energy [64]. After the second baking step, the uncured polymer is removed, leaving only the material forming the negative mold of the desired microchannel (Figure 15-3).

### 4.1.1 Photolithography recipe for SU-8 2075

To fabricate a microchannel mold using SU-8 2075, the wafer was first rinsed with acetone, methanol, and then isopropyl alcohol. It was baked at 115 °C for at least five minutes to evaporate any liquids. The wafer was then placed on a spin coater (WS-400-6NPP-LITE; Laurell Technologies Corp., North Wales, PA, USA), where it was held in place with a vacuum chuck. Approximately 4 mL of SU-8 was poured carefully onto the centre of the wafer and the wafer was spun according to the program in Table 2, forming an approximately 100 μm thick layer of SU-8 on the wafer.

<table>
<thead>
<tr>
<th>Step #</th>
<th>Start speed</th>
<th>Acceleration</th>
<th>End speed</th>
<th>Time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0</td>
<td>110 rpm/s</td>
<td>500 rpm</td>
<td>5 s</td>
</tr>
<tr>
<td>2</td>
<td>500 rpm</td>
<td>--</td>
<td>500 rpm</td>
<td>10 s</td>
</tr>
<tr>
<td>3</td>
<td>500 rpm</td>
<td>330 rpm/s</td>
<td>2000 rpm</td>
<td>5 s</td>
</tr>
<tr>
<td>4</td>
<td>2000 rpm</td>
<td>---</td>
<td>2000 rpm</td>
<td>25 s</td>
</tr>
</tbody>
</table>

In order to obtain sufficient adhesion between the photoresist and the silicon, it was necessary to modify the recommended recipe by increasing the exposure energy and the bake times; the result was that the channels were overexposed and wider than the mask but sufficiently well-adhered to the silicon to be used for multiple PDMS replicas.

After spin-coating, the wafer was soft baked at 65 °C for 5 minutes and then at 95 °C for 60 minutes, rather than the recommended value of 20 minutes. After it cooled, it was selectively
exposed through a mask to 400 nm ultraviolet (UV) light with a Canon PLA-501F mask aligner (Canon Inc., Tokyo, Japan). Exposure was also attempted with the Karl Suss MJB-3 mask aligner in the UBC Cleanroom, but the wavelength of the light provided by that instrument (320 nm) is too low for optimized SU-8 cross-linking [64]. Successful fabrication was achieved with four exposures of 45 s each, with a 45 s pause in between exposures to avoid any excessive heating of the wafer. The exposure energy of the mask aligner was approximately 7 mW/cm². Following exposure, the wafer was post-exposure baked at 65° C for 4 minutes and at 95° C for 15 minutes, which was increased from the recommended 9 minutes.

The next step was the development process which removes the uncured SU-8 material from the wafer. The wafer was immersed in a shallow bath of SU-8 developer (MicroChem Corp., Newton, MA, USA) and the bath was gently agitated for 8-10 minutes until all of the unexposed SU-8 was removed from the wafer. The wafer was then rinsed off with fresh developer and isopropyl alcohol, and blown dry with nitrogen. Lastly, the wafer was hard-baked at 150° C for 5 minutes. This step is helpful in the removal of surface cracks in the SU-8 and promotes adhesion between the SU-8 and the wafer.

SU-8 and silicon have different thermal properties, requiring some caution to be used in the baking steps in the photolithography process, particularly with the SU-8 2075. The thermal expansion coefficient of SU-8 is around 52 ppm K⁻¹, whereas that of silicon is around 3 ppm K⁻¹ [65]. If the temperature is changed too rapidly, the resulting mismatch of thermal expansion at the interface of the two materials can lead to poor adhesion between them. If there is poor adhesion, the SU-8 can delaminate from the wafer during development, or later when the wafer/SU-8 mold is being used to form channels. To minimize this thermal damage, the temperature of the hotplates and wafer is raised at approximately 10°C/min in all baking steps, and at the end of the baking processes, the hotplate was set to room temperature and allowed to cool to below 40° C at approximately 4°C/min before the wafer is removed. For channels with widths below 40 µm, the heating ramp rate was changed to 5°C/min.

4.1.2 Photolithography with SU-8 3050

The SU-8 3050 photoresist was found to have significantly better adhesion between the photoresist and the wafer. The manufacturer-recommended process could be used with very little modification, resulting in channels that were actually the width of the mask and not overexposed as the SU-8 2075 channels were.
Figure 16. Images of the SU-8 mold on silicon wafer. In (a), the circle (diameter: 2 mm) will be a target for punching a hole to access the sheath air inlet channel; the sheath air inlet channel, seen on the bottom left of the image, is 100 μm high by 40 μm wide. In (b), the node where the two sheath inlets and sample inlet join together is seen at the bottom right of the image, with the hole punch target for the sample inlet seen at the top left of the image. The sheath and sample inlets are 100 μm high by 40 μm wide, and the main separation channel after they join is 100 μm high by 60 μm wide. Both images were taken on a stereo microscope (SZ61 Zoom Stereo Microscope, Olympus Inc.).

The viscosity of the SU-8 3050 is lower than that of the 2075, so a lower spin speed had to be used. The spin program shown in Table 3 was recommended by the manufacturer for a height of 100 μm; using this program resulted in channels that were between 97 μm and 102 μm high (Figure 16).
Table 3. Spin program used with SU-8 3050 for 100 µm height.

<table>
<thead>
<tr>
<th>Step #</th>
<th>Start speed</th>
<th>Acceleration</th>
<th>End speed</th>
<th>Time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0</td>
<td>110 rpm/s</td>
<td>500 rpm</td>
<td>5 s</td>
</tr>
<tr>
<td>2</td>
<td>500 rpm</td>
<td>--</td>
<td>500 rpm</td>
<td>10 s</td>
</tr>
<tr>
<td>3</td>
<td>500 rpm</td>
<td>330 rpm/s</td>
<td>1500 rpm</td>
<td>3 s</td>
</tr>
<tr>
<td>4</td>
<td>1500 rpm</td>
<td>---</td>
<td>1500 rpm</td>
<td>30 s</td>
</tr>
<tr>
<td>5</td>
<td>1500 rpm</td>
<td>1650 rpm/s</td>
<td>0 rpm</td>
<td>1 s</td>
</tr>
</tbody>
</table>

To use the SU-8 3050, the general recipe described for SU-8 2075 above was followed, but with the baking and exposure parameters listed in Table 4.

Table 4. Parameters used for SU-8 3050 fabrication process.

<table>
<thead>
<tr>
<th>Step</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Soft bake</td>
<td>50 min. @ 95°C</td>
</tr>
<tr>
<td>Exposure</td>
<td>67 s @ 7 mW/cm² energy</td>
</tr>
<tr>
<td>Post-exposure bake</td>
<td>1 min. @ 65°C</td>
</tr>
<tr>
<td></td>
<td>4 min. @ 95°C</td>
</tr>
<tr>
<td></td>
<td>1 min. @ 65°C</td>
</tr>
</tbody>
</table>

The temperature ramping used for the SU-8 2075 photoresist was not necessary for the SU-8 3050; the one minute long intermediate bake at 65°C before and after the 95 °C post-exposure bake was sufficient.

4.1.3 Implications of photolithography process for mask design

The wafers produced in the photolithography step are used as negative molds for microchannel formation. SU-8 is a negative photoresist, meaning that it becomes cross-linked and remains adhered to the silicon wafer substrate in the areas that were exposed to the UV light. The implication of this property is that when a mask is created through which to expose the wafer, the shape of the channel must be transparent, and all other areas of the mask must be opaque (Figure 17).
Making channels

The separation channels were made of PDMS (Sylgard® 184, Dow Corning), a two-component silicone polymer. The PDMS was combined in a ratio of 10 parts base to 1 part hardener and mixed by hand for approximately 30 s. A Thinky Planetary Centrifugal Mixer (Thinky USA, Laguna Hills, CA, USA) was then used to further mix the PDMS, for 2.5 min. on the “mix” setting and 1.5 min. on the “degas” setting. Approximately 15 mL of the PDMS was then carefully poured onto the wafer mold (Figure 18a), which had been placed on an aluminum foil “boat” which provided sidewalls to contain the PDMS (“Piece A”, Figure 15 and Figure 32). Another 10 mL of PDMS was poured onto a clean, unstructured glass microscope slide, also with aluminum foil walls, and to be used to form the fourth channel wall (“Piece B”,

Figure 17. (a) Section of the optical mask design and (b); image of the resulting SU-8-on-silicon mold; image is taken with a WYCO profilometer. The three inlet channels are 40 μm wide and 100 μm high; the main channel section, after the three inlets meet, is 60 μm wide by 100 μm high.
Figure 15 and Figure 32). The PDMS was degassed in a vacuum dessicator and both pieces were cured in an oven at 80°C for 2.5 hours and then peeled off from their respective substrates. Holes were punched through Piece A using a 0.75 mm borer (Harris Unicore) to provide access to the channel inlets.

Initially, the holes were punched vertically in the centre of the sample air inlet hoe-punch region. However, it was observed that particles, particularly those over 1 µm in diameter, were depositing in the large inlet region before actually reaching the inlet channel. To determine
whether this was an inertial impaction effect, holes were punched at an angle to the vertical to decrease the angle of change of the air streamlines. However, this deposition was still observed, and the holes punched at an angle had high rates of leakage from small tears in the PDMS along the hole. It was thus determined that the deposition may be due to gravitational settling, so for any further devices, the holes were punched vertically as close to the entrance of the inlet channel as possible.

After the holes were punched in Piece A of the PDMS, both pieces were treated with an oxygen plasma for 45 s (Expanded Plasma Cleaner on “high” setting, from Harrick Plasma, Ithaca, NY, USA) and the flat side of Piece B was pressed against the channel side of Piece A. The bonded PDMS pieces were placed in an oven at 80° C and pressed together with a 100 g mass for 1-2 hours to ensure bonding.

After the bonding was complete, the PDMS was transected through the channel near the outlet to allow a filter to be placed into the path of the particles in the channel during experiments (Figure 18b).
5. EXPERIMENTAL SETUP AND PROCEDURES

5.1 Test aerosols

Polystyrene latex (PSL) microspheres were used as test particles. PSL microspheres are a common sample in aerosol research, as they are available in a wide range of highly uniform sizes and with a variety of dyes. The PSL microspheres are suspended in water when purchased; their surfaces are carboxyl-functionalized, causing them to be hydrophilic which minimizes agglomeration. With a density of 1.050 g/mL, the microspheres are nearly neutrally buoyant. PSL microspheres are available that are fluorescently-dyed to increase their visibility when viewed under an epifluorescence microscope; the different dyes allowed for the identification of different sizes of particle. We used the dyes and sizes shown in Table 5.

<table>
<thead>
<tr>
<th>Dye</th>
<th>Dye shorthand</th>
<th>Manufacturer</th>
<th>Particle sizes [μm]</th>
<th>Max. excitation [nm]</th>
<th>Max. emission [nm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>yellow green</td>
<td>YG</td>
<td>PolySciences</td>
<td>0.75 2.88</td>
<td>441</td>
<td>486</td>
</tr>
<tr>
<td>suncoast yellow</td>
<td>SY</td>
<td>Bangs Labs</td>
<td>0.97 2.28</td>
<td>540</td>
<td>600</td>
</tr>
<tr>
<td>orange</td>
<td>O</td>
<td>Invitrogen</td>
<td>0.2</td>
<td>540</td>
<td>560</td>
</tr>
<tr>
<td>green</td>
<td>TG</td>
<td>Thermo Scientific</td>
<td>1.9</td>
<td>468</td>
<td>508</td>
</tr>
<tr>
<td>red</td>
<td>TR</td>
<td>Thermo Scientific</td>
<td>1.0 3.2</td>
<td>542</td>
<td>612</td>
</tr>
<tr>
<td>blue</td>
<td>TB</td>
<td>Thermo Scientific</td>
<td>1.0</td>
<td>412</td>
<td>445</td>
</tr>
</tbody>
</table>

5.2 Microscope

A Nikon Eclipse TE2000-U inverted microscope was used to image the results of the experiments. A LaVision black and white camera (Imager ProX, LaVision GmbH) was attached to the microscope and feeds images directly to a PC for capture. Images were captured using DaVis (version 7.2, LaVision GmbH).

The microscope has three sets of epi-fluorescence interference and absorption optical filters. Each filter set has one (“excitation”) filter allowing light to illuminate the sample, one dichroic mirror, and a second (“emission”) filter allowing light to pass from the sample to the
eyepiece or camera. The four optical filters available on the microscope are described in Table 6.

**Table 6. Optical filter characteristics.**

<table>
<thead>
<tr>
<th>Filter</th>
<th>Shorthand</th>
<th>Excitation centre (bandwidth) [nm]</th>
<th>Mirror [nm]</th>
<th>Emission centre (bandwidth) [nm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>31020</td>
<td>3</td>
<td>480 (30)</td>
<td>505</td>
<td>605 (55)</td>
</tr>
<tr>
<td>41001</td>
<td>GG</td>
<td>480 (40)</td>
<td>505</td>
<td>535 (50)</td>
</tr>
<tr>
<td>Z532LP</td>
<td>LZ</td>
<td>532 (10)</td>
<td>545</td>
<td>n/a – high-pass emission above 550</td>
</tr>
<tr>
<td>49000DT</td>
<td>B</td>
<td>460 (25)</td>
<td>410</td>
<td>350 (25)</td>
</tr>
</tbody>
</table>

To successfully image fluorescent particles, it is necessary that the wavelengths at which the particle’s fluorescent dye is excited lay in the range of wavelengths passed by the illumination filter, and that the wavelength of the light emitted by the dye lay in the range of wavelengths passed by the emission filter. Images were taken with the microscope and camera to verify the colour and fluorescence of the purchased particles. The different colours allow different sizes of particles to be distinguished from one another. This is illustrated with Figure 19, which shows two different particles, with different dyes, under an optical microscope with two different filters. With the “LZ” filter, the particles with suncoast yellow dye are clearly visible, but when the filter “3” is used instead, the particles do not show up at all. On the other hand, the yellow-green particles are clearly visible under the “3” filter but not at all under “LZ”.


5.3 Aerosol generation

To aerosolize the PSL microspheres for use as airborne test particles, a nebulizer was purchased and attached to a custom built drying column. A nebulizer is a device which makes a liquid into fine droplets to create a mist; nebulizers are commonly used for administering liquid medication [66][67]. There are two main types of nebulizers used for generating a PSL aerosol: air jet nebulizers, where energy from a flow of compressed gas is used to aerosolize the liquid, and ultrasonic nebulizers, where energy from a vibrating surface is coupled to the liquid and causes small droplets to form from the surface [68]. Droplets produced by ultrasonic nebulizers typically have less momentum than those produced by air jet nebulization [69]. This can be an advantage when using the technology to uniformly coat surfaces, which nebulizers are sometimes used for, but can lead to a higher rate of deposition due to settling, particularly in low flow rate conditions [70]. Ultrasonic nebulizers cause the sample liquid to heat up, which can lead to higher rates of undesired particle aggregation.

Aerosols can also be generated by electrospray systems. Electrospray atomization creates small droplets of a conductive liquid by applying a high electric field at the outlet of a
capillary tube. A balance of electrical and surface tension forces causes the formation of a thin jet of fluid which breaks apart to form small droplets [71]. This technique has been shown to produce a highly monodisperse aerosol [72], but it has several disadvantages. The chief disadvantage, in light of the requirements for the present work, is that electrospray atomization generally produces droplets in the range of a few nanometers up to a few hundred nanometers. The TSI Electrospray Aerosol Generator (Model 3480 Electrospray Aerosol Generator, TSI Inc.), for example, has a stated range of 2-100 nm. Furthermore, the liquid has to be electrically conductive and the pH carefully controlled, as the mean droplet size depends on the conductivity of the liquid, and the resulting droplet is highly charged, and therefore needs to be electrically neutralized before it can be used with any other instruments; high losses due to electrostatic deposition can still occur [73].

Mahurin and Cheng [72] studied the generation of nanoparticles with air jet, ultrasonic, and electrospray nebulizers and observed that the air jet nebulizers provided a simple and easy method for generating particles, but that they had a wider range of droplet sizes than the other generation methods. Air jet nebulizers characteristically have a significantly higher flow rate output than is necessary for microfluidics. However, given the disadvantages of the other approaches, the relative simplicity of setting up an air jet nebulizer system, and the fact that such a system was available for initial testing and use, an air jet nebulizer system was used for this project.

5.3.1 Vertical drying column

The nebulizer and original drying assembly shown in Figure 20 were based on a similar setup in Dr. Steven Rogak’s laboratory in Mechanical Engineering. The output of the nebulizer was coupled to a tall, vertical drying column, which absorbed the humidity in the air and any water droplets produced by the nebulizer, leaving behind a dry aerosol in dehumidified air. This drying tube was made of a 6 ft. long clear plexiglass tube, 3 inches in diameter. PVC pipe fittings were put on the top and bottom, and clear plastic endplates were machined to be fixed onto the PVC fittings to create ports for the wet aerosol inlet and dry aerosol outlet. The whole assembly was sealed with silicone sealant. Inside the plexiglass tube is an inner tube of permeable mesh made from aluminum screen-door material wrapped with heavier chicken wire for mechanical support. Dessicant was poured between the inner mesh tube and outer plexiglass tube to create a diffusion dryer; the dessicant used was indicating Drierite (CaSO₄; W. A.
Hammond DRIERITE Co. LTD, Xenia, OH, USA), which changes colour from light blue to purple as it absorbs moisture.

Copper tubing (3/8” outer diameter) was used to connect the nebulizer to the drying tube. The drying tube was mounted onto a lab coat rack to stand it up vertically; a bracket was mounted onto the rack to support the tube’s weight and the a fixtures was added to support the nebulizer cup assembly.

Figure 20. Original nebulizer and drying column assembly. As pictured, the pressurized air for the nebulizer was provided by compressed nitrogen tanks. This assembly was later moved and an in-house compressed air line used instead.
5.3.2 Replacement of the vertical drying column with a smaller horizontal dryer

The vertical drying column was used for initial experiments, but it was found that a large quantity of dust was generated by the movement of the Drierite dessicant. This pale blue-purple dust was visible in the tubing and needle connectors to the microchannel, particularly if the setup had been moved just before. Additionally, the setup was difficult to take apart to remove the dessicant to dry it. Thus, this drying column was replaced with a smaller drying tube, designed to be easily pulled apart and to generate a minimal amount of dust. The result was a small horizontal tube, with silica gel (Silica Gel Tel-Tale® Desiccant, Indicating, 10-18 Mesh, Fisher Scientific Inc.) lining the bottom of the tube (Figure 21).

The main disadvantage of this dryer design is the potential loss of particles due to gravitational settling while in residence over the silica gel. Equating the gravitational force with the viscous drag force

\[ F_{\text{Stokes}} = F_{\text{gravity}} \]
\[ 3\pi \mu V d = \frac{4}{3} \pi r^3 \rho g \]

provides the settling velocity

\[ V_{\text{settling}} = \frac{d^2 \rho g}{18 \mu} \]

of a particle under the influence of gravity.

Knowing the typical flow rate \( Q_{\text{air}} \) of air from the nebulizer (see next section) and the cross-sectional area \( A_{\text{tube}} \) of the drying tube, the mean air velocity can be calculated. A 1” tube was used, and it was assumed that it would be one-third filled with silica gel. This provides an approximation to the average residence time of a particle in the tube, and thus the average vertical distance

\[ d_{\text{settled}} = V_{\text{settling}} \times L_{\text{tube}} \times \frac{A_{\text{tube}}}{Q_{\text{air}}} \]

that a particle would fall while in the tube as a function of the tube length \( L_{\text{tube}} \) (Figure 22).

It was decided to use a tube length of approximately 15 cm, because it was found that this length could hold sufficient silica gel to dry nebulized sample air for approximately 45 minutes, which was a sufficiently long time to run a typical test. Both the inlet and outlet were around 5 mm in diameter, so a particle having a 1-2 mm settling distance was acceptable.
Figure 21. Nebulizer assembly with horizontal drying column.

Figure 22. Distance settled by a 3.2 μm particle in the drying tube as a function of tube length, with a 0.8 L/min flow rate.
5.3.3 Initial nebulizer characterization

Several characterizations of the nebulizer were performed, using the vertical drying column and an in-house compressed air line in the ICICS Building at UBC as the pressurized air source. The flow rate was measured as a function of the compressed air pressure using a flow meter (TSI Flow Meter Model 4199) at the outlet of the drying column (Figure 23).

Measurements of the particle concentration produced by the nebulizer under various conditions were taken with the APC-1000 particle counter (Biotest Diagnostics, Denville, NJ, USA). The APC-1000 is an optical particle counter; it has a pump for active sampling at 0.1 ft³/min. The APC-1000 is not able to sample under an imposed pressure difference between the sample air and its own air outlet, so the nebulizer output air stream was only loosely coupled to the APC-1000 to avoid applying such a pressure difference. The APC counts particles in four different size bins: larger than 0.3 μm, 0.5 μm, 1.0 μm, and 5.0 μm diameter.

![Figure 23. Flow rate of air leaving the nebulizer assembly as a function of nebulizer jet air pressure.](image)

The nebulizer jet head has a set screw that acts as a primary baffle, and its position affects the output of the nebulizer. To optimize the screw position, measurements of the nebulizer output as a function of screw position were taken with the APC-1000, with deionized water as the sample, and with the APC measuring the background particle count from the nebulization of the deionized water. When the screw is fully in (0 turns) or fully out (8 turns), very little nebulization of the liquid occurs. The tests revealed that the most particles were generated when
the screw was two full 360° turns from the fully-in position, but that any number of screw turns between the two extremes still allowed some nebulization to occur (Figure 24).

![Figure 24. Concentration of nebulized particles of diameter d as a function of set screw position](image)

**5.3.4 Dependence of nebulization rate on particle size**

Past literature has reported a size-dependent nebulization rate for polystyrene latex microspheres in other air-jet nebulizers [68]. To calibrate the setup for this phenomenon, solutions were prepared such that each had an equal number of microspheres of two different sizes (Table 7). The sample was nebulized and the aerosolized particles were caught on a MetriCEL 0.8 μm pore filter (PallFlex E70-2075W, Pall Corporation) before or after the drying column. The filter was held in a plastic sampling cassette, a three-piece plastic device which holds the filter tightly in place against an outlet with grooves behind it to allow air flow while still supporting the filter. Each combination of particles was tested on each end of the drying column three separate times, each time with a new filter.

**Table 7. Combinations of particle sizes used for characterization of nebulizer size-dependency tests.**

<table>
<thead>
<tr>
<th>Test</th>
<th>Particle 1</th>
<th>Particle 2</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.75 μm</td>
<td>1 μm</td>
</tr>
<tr>
<td>2</td>
<td>1 μm</td>
<td>1.9 μm</td>
</tr>
<tr>
<td>3</td>
<td>1 μm</td>
<td>2.28 μm</td>
</tr>
<tr>
<td>4</td>
<td>2.28 μm</td>
<td>2.88 μm</td>
</tr>
</tbody>
</table>
The optical microscope was used to take images of each filter in three separate locations on each filter, with the optical filters being used to distinguish between the differently sized and dyed particles at each location (see example, Figure 25). After the images were recorded, they were analyzed in MATLAB using the bwboundaries function to identify the bright spots on the dark background. The MATLAB code’s ability to accurately count the number of particles was confirmed by comparing its result with manually-counted particle numbers using small sections of some of the images.

![Image 25](image.png)

Figure 25. 0.75 um (a) and 2.28 um (b) particles after nebulization of a sample containing equal numbers of each size; the images are at the same location on the MCE filter, taken with different optical filters.

The proportion of the number of the two particle types for each set of pictures was found and averaged over all the images taken for each size combination. The final results were normalized so that the concentration of 0.75 μm beads produced equaled one and all other counts were relative to that. There was a clear correlation between larger particle diameter and decreased nebulization efficiency (Figure 26); the final numbers were used as correction factors in later analyses.

The measurements were taken both before and after the drying column to determine whether the size-dependent decrease in concentration in the final output was a result of the nebulizer jet assembly itself or whether the drying column and connections were contributing to the losses. Aside from gravity, the two main effects which could cause loss in the drying column are electrostatic attraction forces on charged particles and inertial forces causing particles to impact and remain on surfaces in the column or attached connectors or tubing. It is expected that smaller particles would be more affected by electrostatic forces and larger particles by inertial
forces. To avoid the buildup of electrostatic charges, conductive or antistatic tubing was used as much as possible in the connections to and from the drying column. To prevent inertial forces from causing particles to impact on surfaces, the curved copper tubing connecting the nebulizer jet assembly to the drying column had a large (>20 cm) radius of curvature, and the connections between the outlet of the drying column and the filter for collecting samples were kept straight. Later, when a microchannel was attached, it was necessary to have a 90° bend in the tubing direction, and to minimize inertial impacts this was done by curving the tubing with a radius of curvature of a few cm. Potential gravitational losses were primarily addressed by the drying column design (section 5.5.1). A slight difference was seen after the drying column for the two larger bead sizes, but in both cases that difference was within the range of variation in the tests. It is therefore concluded that while there may be some inertial and electrostatic losses in the drying column, any size-dependent losses in the drying column are not significant relative to the size-dependent losses from the nebulizer.

Figure 26. Nebulization efficiency as a function of particle size
5.3.5 **Background and particle generation measurements**

Further characterization was performed with the nebulizer to measure the particle levels generated by nebulizing samples of deionized (DI) water, distilled water, and water with 0.545 μm microspheres in concentrations ranging from 1 ppb to 1000 ppb. The water samples examined the background particle levels present in the system, and the samples with microspheres were used to establish what concentration of particles was necessary to sufficiently overcome those background levels. The APC-1000 was used to measure all particle counts.

Both the deionized (Figure 27) and distilled (Figure 28) water samples were compared to particulate levels commonly found in indoor and outdoor environments and to those found with other nebulizer tests [74]. To compare the measured number of particles/ft$^3$ of air to the more commonly used units of μg/m$^3$, it was assumed that the particles in each size bin were the middle of the bin’s range (e.g., that all of the particles in the 0.3 μm – 0.5 μm bin were 0.4 μm, etc.) and that the particle density was 1000 kg/m$^3$. This yielded a result of 5.4x10$^{-4}$ μg/m$^3$ ± 1.1x10$^{-4}$ μg/m$^3$ for the deionized water and 6.9x10$^{-4}$ μg/m$^3$ ± 2.8x10$^{-4}$ μg/m$^3$ for the distilled water. However, the APC flow rate of 0.1 ft$^3$/min = 2.8 L/min was 3.5 times higher than the nebulizer output of 0.8 L/min at the pressure applied. Applying this dilution factor to the results yields a background level of 1.9x10$^{-3}$ μg/m$^3$ ± 3.9x10$^{-4}$ μg/m$^3$ for the deionized water and 2.4x10$^{-3}$ μg/m$^3$ ± 0.96x10$^{-3}$ μg/m$^3$ for the distilled water. Studies of outdoor ambient air typically result in particle ranges from 10$^{-4}$-40 μg/m$^3$ [75][76]. After correcting for dilution, the number count of particles from the nebulization of DI and distilled water was on average 1.4x10$^6$ particles/ft$^3$ air. Sioutas [74] found that nebulized indoor air contained 10$^4$ particles/cm$^3$ air, and nebulized deionized water contained 0.3x10$^3$ particles/cm$^3$ air; this is equal to roughly 3x10$^8$ and 8x10$^6$ particles/ft$^3$ air, respectively. Clifford *et al.* [77] used a pneumatic nebulizer to measure particle concentration in argon gas and deionized water and found particle concentrations of approximately 5x10$^6$/ft$^3$ in the gas and 6x10$^9$/ft$^3$ from the DI water, with the particle diameters less than 0.5 μm.

Smaller particles had significantly higher concentrations, with the particle count (number per volume of air) approximately one order of magnitude higher for each smaller size bin of the APC. Tests with 0.545 μm PSL particles showed a good correlation between the sample concentration and the resulting aerosol concentration (Figure 29). However, dilutions of 1 ppb and 10 ppb particles were too low to distinguish from the background readings from DI water (Figure 30) and distilled water (Figure 31). For separation and detection experiments,
concentrations higher than 1000 ppb were used but they were not included in these tests because the upper limit measurable by the APC is $10^6$ particles/ft$^3$.

Figure 27. Particles nebulized from deionized water, uncorrected for dilution.

Figure 28. Particles nebulized from distilled water, uncorrected for dilution.
Figure 29. Concentration of aerosolized particles as a function of concentration of particles in solution before nebulization, uncorrected for dilution. The particle diameter is 0.545 µm. Only data from the 0.5-1.0 µm bin are shown.

Figure 30. PSL particles (diameter: 0.545 µm) nebulized from various concentrations of particles diluted in DI water, compared to the background reading from DI water. Only the data from the 0.5 µm – 1.0 µm bin of the APC is shown; data is uncorrected for dilution.
Two types of channels were fabricated for experiments. The first were straight channels (Figure 32), used to verify the flow resistance of the channels and the sheath to sample ratio resulting from various inlet pressures, and to observe the cross-streamline diffusion of particles. The channels dimensions are listed in Table 8.

**Table 8. Straight channel dimensions.**

<table>
<thead>
<tr>
<th>Feature</th>
<th>Dimension</th>
<th>Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sheath air inlets</td>
<td>length</td>
<td>5 cm</td>
</tr>
<tr>
<td>Sheath air inlets</td>
<td>width</td>
<td>40 μm</td>
</tr>
<tr>
<td>Sample air inlet</td>
<td>length</td>
<td>0.3 cm</td>
</tr>
<tr>
<td>Sample air inlet</td>
<td>width</td>
<td>40 μm</td>
</tr>
<tr>
<td>Outlet</td>
<td>length</td>
<td>4 cm</td>
</tr>
<tr>
<td>Outlet</td>
<td>width</td>
<td>60 μm</td>
</tr>
<tr>
<td>All</td>
<td>height</td>
<td>100 μm</td>
</tr>
</tbody>
</table>

Figure 31. PSL particles (diameter: 0.545 μm) nebulized from various concentrations of particles diluted in distilled water, compared to the background reading from distilled water. Only the data from the 0.5 μm – 1.0 μm bin of the APC is shown; uncorrected for dilution.

5.4 Straight channel experiments

Two types of channels were fabricated for experiments. The first were straight channels (Figure 32), used to verify the flow resistance of the channels and the sheath to sample ratio resulting from various inlet pressures, and to observe the cross-streamline diffusion of particles. The channels dimensions are listed in Table 8.
5.4.1 Measuring flow resistance

To confirm the flow resistances of the straight channels, a syringe pump (KD Scientific) was used to push air through the channels at a known flow rate $Q$, while the pressure at various points of the channel was measured with pressure sensors (ASDX Series 0-1 PSI and 0-15 PSI ranges, Honeywell). The ASDX sensors output an analog voltage signal $V_{out}$ proportional to the differential pressure between the two inlet ports,

$$P = (V_{out} - 0.5V) \left( \frac{P_N}{4V} \right),$$

where $P_N$ is the nominal range of the pressure sensor (in this case, 1 PSI or 15 PSI).

Two setups were used for measuring the flow resistance, as shown in Figure 33, where the channels are represented with resistors in an analogy to Ohm’s Law. In the first setup (Figure 33a), the syringe pump flow was connected to the two sheath inlets. The pressure was measured at the sheath inlets and also at the sample inlet. Since there is no flow travelling through the sample inlet, the pressure measured there is equal to the pressure at the node $P_C$. The difference between the pressure at the sheath inlets and $P_C$, divided by the flow rate, gives the total resistance of the two parallel sheath inlets. The pressure drop $P_C$ across the outlet channel divided by the flow rate gives the flow resistance $R_4$ of that section. The second setup (Figure 33b) was used to measure the sample inlet flow resistance. The setup was similar to the first one,
but the flow entered the sample inlet instead of the sheath inlets, and the pressure $P_C$ was measured through the sheath inlets.

Tests were done with and without a filter in place at the channel exit and at several flow rates between 100 $\mu$L/min and 1500 $\mu$L/min. The flow resistances were

$$R_2 = R_3 = \frac{2 (P_{\text{sheath inlet}} - P_C)}{Q}$$  \hspace{1cm} \text{(Figure 33 a)}

$$R_1 = \frac{P_{\text{sample inlet}} - P_C}{Q}$$  \hspace{1cm} \text{(Figure 33 b)}

$$R_4 = \frac{P_C}{Q}$$  \hspace{1cm} \text{(Figure 33 a, b)}.

---

Figure 33. Arrangements used for measuring flow resistance of the straight and curved channels, with the imposed flow from the syringe pump labeled $Q$ and the pressure sensors marked with an encircled “$P$”.
These flow resistance values were used with Equations 7-10 to determine parameters for later experiments and analysis. Every time a new mold was made, the first setup was modified slightly once to test that the two sheath inlets were equal in resistance to another by inputting the flow into each sheath inlet separately and blocking the other sheath inlet. If the two inlets were of the same resistance, this was not repeated for each channel replicated from that same mold, but the flow resistance of the combined sheath inlets, sample inlet, and outlet section were still measured for each channel to monitor for any problems with the mold replica process and particularly also to identify any leaks from the inlet holes.

5.4.2 Testing channel and flow characteristics with airborne particles

Experiments with the straight channel using airborne fluorescent polystyrene latex microspheres were undertaken to verify the sheath to sample ratio as a function of the inlet pressures, to examine the effects of diffusion, and, more generally, to test and improve the experimental setup. The channels had a rectangular piece of 0.8 μm pore mixed cellulose ester filter (GN-4 MetriCEL filter, Pall Corporation) placed in the cut at the channel exit, and the channel was sealed around the filter and held tightly with a clamp (Figure 34 and Figure 35). The clamp consisted of two pieces of plastic held in parallel; one had a rectangular hole cut in the centre to allow the air leaving the channel to flow out.

Figure 34. Straight channel in clamp.
To determine whether the filter choice was suitable, given the flow rates used and the small area through with the flow had to travel, a rough comparison with the approved regulatory methods was made. The US National Institute for Occupational Safety and Health (NIOSH) publishes methods for collecting and analyzing a wide variety of hazardous materials, including particulates. NIOSH method number 7300 describes the collection of airborne metal particulates by use of a mixed cellulose filter with 0.8 µm pore size [78]. Using a 37 mm Pall GN-4 filter, which is approved for this method, the range of flow rates per area of filter that could result from the use of this method is between approximately 0.9 and 3.7 mL/min/mm$^2$. The flow rates per area of filter used with the microchannel were between 0.3 and 3.3 mL/min/mm$^2$. The NIOSH method is recommended for sampling metal particulates with exposure limits ranging from 2 µg/m$^3$ (arsenic) to 10 mg/m$^3$ (aluminum dust). The estimated mass concentration range of the nebulized polystyrene latex microspheres is around 1 mg/m$^3$ to 40 mg/m$^3$.

The pressure applied at the sample inlet was controlled by a valve connected to a bleed air line at the outlet of the nebulizer, as shown schematically in Figure 36. The pressure was measured and displayed in real time using LabView (version 8.5; National Instruments Corp., Austin, Texas, USA).
Pressure-controlled sheath air was provided by a MFCS-8C microfluidic pressure control system (Flui gent, Paris, France) with a range of 0-1000 mbar and a resolution of 0.3 mbar. For these and the following curved channel tests, each test was allowed to run for between 25 and 120 minutes (depending on the flow rates and particle concentrations), and then the filter was removed and imaged on the microscope.

5.5 Curved channel experiments

Channels with an outlet curved for separation were also made (Figure 18), with the dimensions as shown in Table 9.

5.5.1 Measuring flow resistances

Flow resistance measurements with a syringe pump and pressure sensors, identical to those described in Section 5.4.1, were undertaken with the curved channels as well, and were also used to determine the relationship in experiments between average velocity and applied inlet pressures. The same configurations as those described in Figure 33 were used.
Table 9. Curved channel dimensions.

<table>
<thead>
<tr>
<th>Feature</th>
<th>Dimension</th>
<th>Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sheath air inlets</td>
<td>length</td>
<td>5 cm</td>
</tr>
<tr>
<td>Sheath air inlets</td>
<td>width</td>
<td>40 μm</td>
</tr>
<tr>
<td>Sample air inlet</td>
<td>length</td>
<td>0.3 cm</td>
</tr>
<tr>
<td>Sample air inlet</td>
<td>width</td>
<td>40 μm</td>
</tr>
<tr>
<td>Outlet</td>
<td>length</td>
<td>4 cm</td>
</tr>
<tr>
<td>Outlet</td>
<td>width</td>
<td>60 μm</td>
</tr>
<tr>
<td>Outlet</td>
<td>radius of curvature</td>
<td>5 mm</td>
</tr>
<tr>
<td>All</td>
<td>height</td>
<td>100 μm</td>
</tr>
</tbody>
</table>

5.5.2 Testing separation with single-outlet channels

Initial separation tests were performed with a curved channel with a single widened outlet (Figure 37). As in the straight channel tests, the sample air pressure was controlled with the bleed air valve at the bottom of the nebulizer with a pressure sensor measuring the input to the channel, and sheath air was pressure-controlled by the MFCS-8C from Fluigent. Tests were done on the combinations of particles listed in Table 10, chosen because their different dyes allowed them to be easily distinguishable from each other with the microscope optical filters.

Table 10. Microsphere combinations tested for separation.

<table>
<thead>
<tr>
<th>Combination #</th>
<th>Microsphere 1</th>
<th>Microsphere 2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Diameter (μm)</td>
<td>Dye</td>
</tr>
<tr>
<td>1</td>
<td>0.2</td>
<td>O</td>
</tr>
<tr>
<td>2</td>
<td>0.75</td>
<td>YG</td>
</tr>
<tr>
<td>3</td>
<td>1.0</td>
<td>TB</td>
</tr>
<tr>
<td>4</td>
<td>1.9</td>
<td>TG</td>
</tr>
<tr>
<td>5</td>
<td>2.28</td>
<td>SY</td>
</tr>
</tbody>
</table>
5.5.3 Separation with two-outlet channels

The combinations listed in Table 10 were used to test separation with two-outlet channels. Two designs were used; in both, the outlet expanded to 2000 μm as in a single-outlet channel, and then split into two channels. The designs were chosen by simulating the separation of a mixture of particles and plotting the sum of two the resulting distributions of particle concentration across the channel width. The position in the width where the sum of the distributions was at a minimum was chosen as the position where the flow should diverge into two channels. The flow profile in the pre-split outlet was used to find the relative volumes of flow traveling down each channel, and these values were used to find the ratio of flow resistances of the two sub-outlets. The ratio of flow resistances were used to find the sub-outlet widths required to split the flow in the desired location. The first design then had the channel bifurcated such that the radially outer channel occupied 25% of the full width. The second design had the radially outwards outlet channel at 10% of the full width (Table 11). The first design was based on simulations of 0.2 μm and 0.75 μm particle separation at a mean velocity of 1.8 m/s, and the second on the separation of 2 μm and 3 μm particles at a mean velocity of 0.24 m/s.
### Table 11. Two-outlet channel design parameters.

<table>
<thead>
<tr>
<th>Design #</th>
<th>Outlet #1 (radially inwards)</th>
<th>Outlet #2 (radially outwards)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Width (μm) (out of 2000 μm)</td>
<td>% of flow in that channel</td>
<td>Corresponding width of flow in main channel (out of 60 μm)</td>
</tr>
<tr>
<td>Design #1</td>
<td>1500</td>
<td>76.7%</td>
</tr>
<tr>
<td></td>
<td>500</td>
<td>23.3%</td>
</tr>
<tr>
<td>Design #2</td>
<td>1800</td>
<td>92.7%</td>
</tr>
<tr>
<td></td>
<td>200</td>
<td>7.3%</td>
</tr>
</tbody>
</table>

### Table 12. Three-outlet channel design parameters.

<table>
<thead>
<tr>
<th>Design #</th>
<th>Outlet #1 (radially inwards)</th>
<th>Outlet #2 (middle)</th>
<th>Outlet #3 (radially outwards)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Width (μm) (out of 200 μm)</td>
<td>% of flow in that channel</td>
<td>Corresponding width of flow in main channel (out of 60 μm)</td>
<td></td>
</tr>
<tr>
<td>Design #1</td>
<td>97 μm</td>
<td>63.9%</td>
<td>36 μm</td>
</tr>
<tr>
<td></td>
<td>59 μm</td>
<td>24.2%</td>
<td>12 μm</td>
</tr>
<tr>
<td></td>
<td>44 μm</td>
<td>11.9%</td>
<td>12 μm</td>
</tr>
<tr>
<td>Design #2</td>
<td>101 μm</td>
<td>68.6%</td>
<td>38.1 μm</td>
</tr>
<tr>
<td></td>
<td>58 μm</td>
<td>22.0%</td>
<td>11.4 μm</td>
</tr>
<tr>
<td></td>
<td>41 μm</td>
<td>9.4%</td>
<td>10.5 μm</td>
</tr>
<tr>
<td>Design #3</td>
<td>106 μm</td>
<td>73.7%</td>
<td>40.5 μm</td>
</tr>
<tr>
<td></td>
<td>56 μm</td>
<td>19.1%</td>
<td>10.5 μm</td>
</tr>
<tr>
<td></td>
<td>38 μm</td>
<td>7.2%</td>
<td>9 μm</td>
</tr>
</tbody>
</table>

5.5.1 Separation with three-outlet channels

Similarly to the two-outlet channel design, three-outlet channels were fabricated, to separate three sizes of particle into three separate outlets. The channel design parameters (Table
were chosen with the separation of these particle sizes in mind, with a similar approach as that used to design the two-outlet channels in the previous section. The average air velocities in the simulations which were used to select the outlet widths were 0.2 m/s, 0.23 m/s, and 0.26 m/s, corresponding to designs 1, 2, and 3, respectively. These values were chosen to implement the separation of a three-size mixture of particles in the 1 μm – 3 μm range. The same designs could be used for other particle size ranges, with the air velocities adjusted accordingly.

5.6 Particle detector

The particle detector uses a Pallflex membrane filter (PallFlex E70-2075W, Pall Corporation) held with washers inside a black airtight housing (Figure 38). Light is provided by an LED (LTL1CHTBK2, Lite-On Inc.) directly upstream of the filter. Downstream of the filter, the light is measured with a photodiode (BS500A, Sharp Microelectronics). The supporting electronics are in a box adjacent to the device itself and include a constant DC power line for the LED and signal conditioning for the photodiode signal (Figure 39). The photodiode signal is amplified 10 times with an operational amplifier (LM741) before being sampled at 1 Hz by a data acquisition card (USB-6212, National Instruments). The data is stored and displayed with a LabView program on a nearby computer.

To test for the signal response as a function of particle size and concentration, the detector’s inlet was connected directly to the output of the nebulizer with the vertical drying column attached. A solution with a known concentration of monodisperse polystyrene latex microspheres was put in the nebulizer sample cup, with the mass of the cup and sample measured before and after the test. The nebulizer was run, and the voltage output of the photodiode recorded, for between 0.5 and 3 hours. This test was repeated for several particle sizes (0.75 μm, 2.28 μm, 2.88 μm) at several pre-nebulization concentrations of solution (0.1% w/v, 0.01% w/v, 0.001% w/v).

The estimated volume of the hollow tube in the middle of the nebulizer drying column and the connected tubing is 1.4 L and the flow rate measured through the PSAP was approximately 0.4 L/min resulting in an estimated clearing time of about 3 minutes. To analyze the data produced in each test, the first 500 points (~8 min.) of the data were removed to avoid errors in the data caused by the slow initial response while the nebulizer cleared. The remaining data was fit to a linear curve using Microsoft Excel and the slope was plotted against one of several variables proportional to the concentration of particles in the sample air.
Figure 38. Particle detection system, with the filter assembly and electronics in an enclosable housing.

Figure 39. Schematic of PSAP circuitry
6. RESULTS AND DISCUSSION

6.1 Summary of experimental conditions and parameters

The two types of channels described above – straight channels, to verify the flow focusing and diffusion effects, and curved channels, to observe separation effects – were used in a variety of experiments the results of which are described throughout this chapter. For both sets of experiments, the channels were first fabricated as described in Section 4.2, and the experiments were set up as described in Section 5.4 (straight channels) and Section 5.5 (curved channels). Table 13 provides an overview of the general sets of experiments performed, including the type of channel used, the purpose of the test, and the flow properties and particle sizes used.

<table>
<thead>
<tr>
<th>Channel design (# outlets)</th>
<th>Intended to measure</th>
<th>Particle sizes</th>
<th>Mean air velocity</th>
<th>Sample width (% of channel width)</th>
<th>Inlet pressures</th>
</tr>
</thead>
<tbody>
<tr>
<td>straight channel (1)</td>
<td>flow focusing</td>
<td>0.2 μm, 0.75 μm</td>
<td>0.36 m/s – 1.8 m/s</td>
<td>14% – 40%</td>
<td>22-200 mbar, 10-100 mbar</td>
</tr>
<tr>
<td>straight channel (1)</td>
<td>diffusion</td>
<td>0.2 μm, 1.0 μm</td>
<td>0.7 m/s – 1.4 m/s</td>
<td>35%</td>
<td>30-100 mbar, 17-58 mbar</td>
</tr>
<tr>
<td>curved channel (1)</td>
<td>separation</td>
<td>0.2 / 0.75 / 0.75 / 1.0 / 1.9 μm, 1.9 / 3.2 μm</td>
<td>0.5 m/s – 1.5 /ms</td>
<td>18%</td>
<td>36-156 mbar, 16-65 mbar</td>
</tr>
<tr>
<td>curved channel (2)</td>
<td>multi-outlet separation</td>
<td>0.2 / 0.75 μm</td>
<td>1.8 m/s</td>
<td>12%</td>
<td>160 mbar, 64 mbar</td>
</tr>
<tr>
<td>curved channel (3)</td>
<td>multi-outlet separation</td>
<td>0.2 / 1.0 / 1.9 μm</td>
<td>0.5 m/s</td>
<td>8%</td>
<td>40, 80 mbar, 29.4-29.8 mbar</td>
</tr>
</tbody>
</table>

The choice of particle sizes for the curved channel separation tests was based on the availability of particle sizes with distinguishable fluorescent dyes, with the sizes, dyes, and corresponding optical filters as listed in Table 5 and Table 6. For the straight channel tests, particles on the lower end of the size range were used, because these were nebulized more efficiently (Section 5.3.4). The diffusion tests in particular were performed with particles of sufficiently different diffusion coefficients as to observe the particle-size-dependent diffusion effects.
6.2 Flow focusing in a straight channel

Aerosol samples were aerodynamically focused in a straight channel and collected on a filter at the outlet of the channel as described in section 5.4.2. Images of the collected particles on the filter were taken with an epifluorescence microscope (section 5.2). The images were cropped to a size slightly larger than the outlet of the channel, with the particle-laden section centered in the image (Figure 41a). Each image was scaled to 100 x 2000 pixels (height by width) and imported into MATLAB (v. R2009a, The MathWorks Inc.) using the imread function, which produces an 100 x 2000 x 3 matrix as an RGB (red/green/blue) version of the image. The rgb2gray function was used to collapse the three-dimensional matrix to a two-dimensional one, with each matrix element representing the brightness of its corresponding image pixel. The two-dimensional matrix was summed along each column, resulting in a one-dimensional array of the intensity of the image across the width of the outlet. A moving average was taken using the filtfilt function and then the values were scaled so that the maximum was 100. To determine the number of points to use for the moving average, several straight channel experimental datasets were chosen at random and moving averages using 20, 50, and 100 points were taken of the data. The data from one such test is shown in Figure 40. The 50-point moving average was chosen because it significantly decreased the noise from individual particles without significantly changing the apparent width of the dataset, and was used for all tests.

The resulting values were used to find the full width at half maximum (FWHM) of the concentration (Figure 41b). The simulation had been programmed such that its output was a two-dimensional matrix with the values of the elements representing the number of simulated particles that landed on that element’s equivalent position in the channel cross-section. These simulated matrices were treated in the same way as those resulting from the imported filter photos and the FWHM of the simulated particle runs were compared to those of the experiments. Experiments were run at a variety of expected widths using 0.75 μm particles, and at several velocities with nearly identical expected widths using 0.2 μm particles. Eleven tests of 0.2 μm particles were run at average velocities ranging from 0.7 m/s to 1.4 m/s, with the ratio of sheath to sample flow rates of 1:1. Seventeen tests were performed with the 0.75 μm particles, at average velocities between 0.5 m/s and 1.8 m/s and at ratios of sheath to sample flow rates between 0.8:1 and 5:1. Additional tests were run with 0.2 μm particles at 0.36 m/s; these data are shown in Figure 40 and Figure 41.
The results showed good correlation between the width expected from modeling and the width from the experiments (Figure 42). The mean residual (measured width – expected width) was 0.9% of the full channel width, and the standard deviation of the measurements about the expected values was 6.6% of the channel width.

Figure 40. Moving averages (MA) of one straight channel test, using a 20-, 50-, and 100-point moving average. The test particles were 0.2 µm in diameter, and the mean air velocity in the channel was 0.36 m/s.
Figure 41. (a) Sample results from a straight channel experiment, with (b) concentration plotted against position for both the experiment and a simulation run with identical conditions. The particles were 0.2 µm in diameter and the channel was 60 µm wide and 100 µm high. The total flow rate was 2.15 µL/s, with an average air velocity of 0.36 m/s, and the ratio of sheath to sample flow rates was 1:1. As marked here, the FWHM (section 6.1), W30, and W70 (section 6.3) values are for the simulation.
Diffusion of particles in a straight channel

To quantify the effects of the cross-stream diffusion of particles in the channel, it was necessary to choose a metric that would quantify the deviation of the particles from the binary distribution they would show if there were no diffusion occurring (grey line, Figure 41b). In plots of the simulated and measured concentrations of particles across the channel width, the effect of the diffusion is seen in the sloped sides of the plot (Figure 41b). To quantify these sloped sidewalls and allow for comparison between tests and between experiments and simulations, first the values of the particle concentration plot were scaled so that the mean value of the centre region of the plot, defined as the region with a width of one half of the width of the stream at the beginning of channel, equaled 100 arbitrary units. Then, the difference of the width of the plot at a height of 30% of the centre region value and the width at a height of 70% of the centre region value was found for each dataset. These correspond to the lines marked “W30” and “W70” respectively on Figure 41b, and their difference is henceforth referred to by the shorthand “W30-W70”.

Figure 42. Straight channel width experiment results. The 0.2 µm particles were tested at average velocities ranging from 0.7 m/s to 1.4 m/s and the 0.75 µm particles were tested at average velocities between 0.5 m/s and 1.8 m/s.
Experiments to measure the W30-W70 in straight channels were undertaken with 0.2 µm particles and 1 µm particles. These values were chosen because their difference in sizes leads to diffusion coefficients that are over an order of magnitude apart; the diffusion coefficient (equation 12) of 0.2 µm particles in room-temperature air is 2.7x10^-10, while that of 1 µm particles is 3.0x10^-11.

A significant factor in comparing the simulated diffusion results to the experimental ones is the number of particles used to obtain each dataset. The PSL samples as purchased were in concentrations based on their weight, rather than the number of particles, per volume of water. As a result of this, and because of the decreasing nebulizer efficiency with increasing particle size, the number of particles generated per time decreased significantly with increasing particle size. The number of 0.2 µm particles was so high as to make them appear for the most part indistinguishable from one another. However, significantly fewer 1.0 µm particles were generated. Taking only into account their difference in sizes, a given weight per volume concentration of 0.2 µm particles would have nearly 1000 times more particles than the same weight per volume concentration of 1.0 µm particles. This effect is exacerbated by the decrease in nebulizer efficiency with increased particle size, and running the 1.0 µm experiments for 1000 times longer than the 0.2 µm experiments to compensate was not feasible. To determine the effect that the lower number of particles would have on the results, the simulation was modified so that the initial particle density was lower. The simulations originally placed a particle every 0.3 µm in the height and width dimensions of the sample air stream; with the modification, a particle was placed every N nodes, where the node spacing was the original 0.3 µm. The simulations were run at sheath to sample ratios of 0.9, 1, and 1.1 by flow rates, at average channel velocities of 0.7 m/s, 1 m/s, and 1.1 m/s to match the experimental data, and at values of N =1, 2, 3, … 25. The total resulting W30-W70 values were plotted against the number of particles simulated. Data from the 0.7 m/s simulations is shown in Figure 43. The variations in the W30-W70 were very small (<10%) when the number of particles generated approached or exceeded 10,000. With decreased particle numbers, though, the variation in W30-W70s found in the simulations and experiments increased dramatically.

Sample data (photos and plots) for some of the diffusion tests can be found in Appendix 2.
The expected diffusion behaviour of the 0.2 µm and 1.0 µm particles is summarized with the solid lines in Figure 44; for both, the W30-W70 decreased with increasing velocity, as the decreased transit time in the channel results in less diffusion. The diffusion of the simulated 0.2 µm particles at the highest velocity was still higher than that of the simulated 1.0 µm particles at the lowest velocity, as the order of magnitude difference in their diffusion coefficients dominates the halving of the average transit time seen over the range of velocities tested (equation 15). The 0.2 µm particle experiments were close to the simulated results, but the 1.0 µm particles showed increasingly large error and variance with decreasing velocity experimentally.

The number of particles collected in each experiment was counted or estimated from the filter images for comparison to these simulation results. For the 0.7 m/s tests, the number of 1.0 µm particles collected experimentally was between 75 and 600; for the 1 m/s tests, between 500 and 1500, and for the 1.4 m/s tests the concentration was too high to count or estimate accurately, so it was taken as being high enough to be in the low-variation region of the simulation. The error bars on the simulation results in Figure 44 represent the maximum and minimum values found in the simulated results for the particle counts corresponding to the
experiments done at the same velocities; they thus represent the range in which we would expect to find the corresponding experimental values.

![Graph](image)

**Figure 44.** Straight channel diffusion experiment results. All simulations and experiments were run at a ratio of sheath to sample flow rates of (1±0.1):1. The dashed error bars are for the simulations, and the solid grey and black error bars are for the experiments.

### 6.4 Separation by centrifugal forces

#### 6.4.1 Testing separation with single-outlet channels

Particle separation experiments in single-outlet channels were undertaken with test PSL particles in the range of 0.2 μm to 1.9 μm in diameter at mean channel air velocities of 0.5 m/s, 1 m/s, and 1.5 m/s. The largest of the particles were only tested at the lowest velocity, as higher velocity tests resulted in the impaction of the particles against the radially outer wall, which results were corroborated by the simulations done of particle deposition shown in section 3.3.1.

For each test, two images were taken of the filter that had been collecting particles in the microchannel outlet, using the two different microscope optical filters corresponding to fluorescence behaviour of the two different particle dyes used. For some of the combinations of fluorescent dyes, there were problems with cross-talk occurring between the two different colours. Specifically, when testing the separation of 0.2 μm (orange) and 0.75 μm (yellow-
green) particles, only the 0.75 μm particles would be visible under the “3” optical filter, but both of the particle sizes would be visible under the “LZ” filter used to image the 0.2 μm particles. A similar effect was also seen when separating the 0.75 μm (yellow-green) particles from the 1 μm (blue) particles. In both cases, the effects of the cross-talk were removed by using the image subtraction feature available in the image recording software. This process was quite straightforward for the former combination, since the positions of the two particle sizes in the channel cross-section were very distinct from one another. This could be confirmed visually for each experiment, since the fluorescent emission colours of the two particles were different enough to be distinguished by eye even though they were viewed with the same optical filter. In the latter case, however, there was more overlap in positions of the two particle sizes because of the smaller size difference. Thus, to avoid inadvertently removing real data from the results, the test time was shortened so that there would be fewer of each particle size on the filter, and therefore less chance of two individual particles overlapping and thus being affected by the image subtraction process. To extract particle position information from the images of particles on the filters, the same process as that described for straight channels (section 6.1) was followed, up to the point of obtaining an array representing the particle concentration across the channel width. This array representing the particle concentration across the channel width was then used to find the mean position of each particle size, by identifying the point in the channel width where there was an equal amount of particles to the radially inner direction as to the radially outer direction. The difference in mean positions for each pair of particle sizes at each velocity was found after repeating each set of experiments at least three times.

A sample separation of 0.2 μm particles and 0.75 μm particles is shown in Figure 45. The distinct positions of the two particle sizes on the filter can be seen; the effects of the low velocity regions at the top and bottom of the channel are also visible as regions of unseparated particles at the top and bottom of the images.

Since the absolute position of the channel outlet was not identifiable on the filter at sufficiently high accuracy, the results of the separation experiments are reported as relative to the position of the 0.2 μm particles at each velocity tested (Figure 46). For mixtures not containing 0.2 μm particles, the position of the larger particle size was taken as its relative position with respect to the smaller particle plus the relative position of the smaller size with respect to 0.2 μm particles under identical experimental conditions. As expected, the mean particle position of larger particles with respect to 0.2 μm particles was further radially outward with increasing particle size and with increasing velocity.
Figure 45. Sample of a single-outlet curved channel separation of 0.2 µm and 0.75 µm particles. The mean channel velocity was 1.5 m/s, and the sheath to sample ratio was 3:1 by flow rates.
The error – the difference between the measured mean particle position and the expected mean particle position – also increased with increasing particle size. This may be due to a similar effect as that seen in the diffusion tests: that the decreasing number of particles generated with increasing particle sizes also increases the error and variation in the tests. While higher concentrations could be used to compensate for this somewhat, it was still impractical for the tests to be run long enough to produce as many >1 μm particles as was easily achieved with the sub-micron particles. This likely contributed significantly to the higher variation and error in the mean values observed with the largest particle sizes here. Despite this, the range of mean particle positions measured for each set of conditions did include the value predicted by the simulations, and the average residuals (measured value – expected value) measured for the

Figure 46. Results of particle separation tests in a single-outlet curved channel, with the resulting mean particle positions reported relative to the mean position of 0.75 μm particles under the same conditions. The simulated results are shown with solid lines; the points represent the mean value of the experimental results, with the error bars representing the maximum and minimum values found. Each set of experimental conditions was tested at least three times, at a ratio of sheath to sample flow rates of 3:1, resulting in a sample stream width at the start of the main channel of approximately 10 μm.
0.5 m/s, 1 m/s, and 1.5 m/s tests were only 3.8%, 1.4%, and -3.0% of the full channel width, respectively (Figure 46).

To characterize the ability of the single-outlet channel to move the particles into distinct size-dependent positions, a separation efficiency $\eta$ was defined so that in addition to the mean particle positions presented in Figure 46, the amount of overlap in particle positions could be quantified. To find this value, the point $x_{sep}$ was defined on the x-axis of the plot of particle concentration across the channel width such that

$$\frac{\sum_{x_{sep}}^{1} \text{large particles}}{\sum_{0}^{x_{sep}} \text{large particles}} = \frac{\sum_{0}^{x_{sep}} \text{small particles}}{\sum_{x_{sep}}^{1} \text{small particles}} = \nu$$

(31)

This value was used to define the separation efficiency

$$\eta = 2 \left[ \left( \frac{\nu}{\nu + 1} \right) - 0.5 \right]$$

(32)

This definition reflects the ability of the system to separate the two particle sizes into separate regions. The percentage of particles of one size which are located in one region is equal to the percentage of particles of the other size located in the other region; this percentage is equal to the ratio $\nu/\nu+1$. The constants in the definition serve to scale the separation efficiency so that two perfectly overlapping distributions of particle concentration would have a separation efficiency of zero, and two distributions with no overlap would have a separation efficiency of one.

Figure 47. Separation efficiency of the single-outlet separation tests. The particle size mixtures were as indicated in the legend, and the definition of separation efficiency is as in Equation 32.
This definition was applied to all of the single-outlet curved channel data shown in Figure 46. The resulting separation efficiencies, shown in Figure 47, clearly mimic the separation behaviour seen in the simulations and mean positions of Figure 46, with the separation efficiency increasing with increasing velocity up to a point at which the particle distributions start to compress against the radially outer wall, leading to a smaller difference in their mean positions and to more overlap in their distributions.

6.4.2 PM2.5 separation tests

Tests to separate particles about the PM2.5 standard were undertaken to examine the device’s ability to separate particles about this standard. This separation was only performed at one velocity (average air velocity in channel: 0.25 m/s), but was repeated three separate times on three different days to ensure its repeatability. The test particles were 1.9 μm and 3.2 μm; the simulations predicted a difference in their mean positions of 12% of the channel width; the experimental results ranged from 9.8% to 16.4 % of the channel width, with an average of 12.1% (Figure 48).

![Figure 48. Results of three separation tests at 0.25 m/s of a mixture of 1.9 μm particles and 3.2 μm particles. This mixture was chosen to represent separation about the PM2.5 regulatory standard. The simulated difference in mean positions is indicated by the dashed horizontal line.]


W = (1940/6000)a+60 μm. This was done to enhance the visibility of the separation for the microscope imaging, but also permitted the gravitational settling of some particles on the floor of the microchannel. This was specifically seen with the PM2.5 measurements, and was confirmed mathematically by calculating the settling velocity of the various sizes of test particle in air, and by combining that velocity with the expanding section geometry to find the average distance settled by a particle traveling through the expanding section (Figure 49). It was clear that particles over 1 μm were expected to deposit on the channel floor if allowed to travel through the whole expanding section.

![Figure 49. Settling distance of a particle traveling through the expanding section of the microchannel outlet, assuming the particle’s axial velocity is the same as that of the fluid’s. The axial distance of the expanding section is 6000 μm, and the 1.9 μm and 3.2 μm particles will have settled to the floor of the channel before reaching the end of the section.](image_url)

Images taken of the microchannel surfaces after a test using 2.88 μm particles confirmed this behaviour (Figure 50). To avoid this problem, the position at which the channel outlet was transected in the expansion was moved further upstream, to a position 525 μm from the start of the expansion, where the width had expanded from the nominal channel width of 60 μm to 230 μm, and that channel was used for the PM2.5 tests described above.
The device described as “Design 1” in Table 11 was fabricated, with a bifurcation splitting the channel outlet into two sub-outlets (Figure 51). The radially outer sub-outlet had a 2.88 μm particles settled in the expanding section of the channel outlet under the influence of gravity.

**Figure 50.** 2.88 μm particles settled in the expanding section of the channel outlet under the influence of gravity.

### 6.4.3 Separation with two-outlet channels

**Figure 51.** Two-outlet curved channel separation system

The device described as “Design 1” in Table 11 was fabricated, with a bifurcation splitting the channel outlet into two sub-outlets (Figure 51). The radially outer sub-outlet had a
width that was one quarter of the full channel width prior to the split. This device was thrice tested with a mixture of 0.2 µm and 0.75 µm particles, with an average air velocity in the channel of 1.8 m/s (corresponding with the velocity used to design this channel, see Section 5.5.3) and a ratio of sheath to sample flow rates of 5:1. The particles were again collected on filter paper in the two sub-outlets and imaged with the epifluorescence microscope, and the relative concentrations of each size in each channel was measured. On average, 86.3% of the 0.75 µm particles were found in the radially outer outlet, while the 0.2 µm particles were nearly evenly split between the two outlets in each experiment (Figure 52).

Figure 52. Two-outlet curved channel separation results

While the preferential transport of the 0.75 µm particles into the radially outwards channel was reasonably successful with this set of experiments, for a complete integrated system it would be important to have the 0.2 µm particles primarily in the radially inwards outlet channel. This performance improvement could best be achieved by using the second design listed in Table 11, which has the split between the two outlet channels moved further radially outwards. The velocity could also be optimized for the separation, but with limited benefits.
Decreasing the velocity would cause the mean position of the 0.2 μm particles to move further radially inward, and thus more towards the radially inwards outlet channel. However, the lower velocity would also mean that the particles will diffuse more while in the channel, counteracting the separation effects.

### 6.4.4 Separation with three-outlet channels

The device listed as “Design 2” in Table 12 was used to test the separation of a mixture of three sizes of particles; the outlet of the channel split into three sub-outlets, with the radially inner channel being the widest and the radially outer channel being the narrowest. A mixture of 0.2 μm, 1.0 μm, and 1.9 μm particles was prepared; each particle size had a different fluorescent dye, distinguishable from the other two sizes, to allow for the independent identification of each particle size.

![Graph](image)

**Figure 53. Three-outlet curved channel separation results.** Five tests were repeated, with an average channel velocity of 0.52 m/s, and a ratio of sheath to sample air flow rates of 9.5±0.7 : 1. The sample stream sheath air inlets were at nonequal pressures, such that before the bend, the sample stream was centred closer to the radially inner wall than the radially outer one, at a position 26% of the way across the channel width. The bars show the average values, with the error bars representing the maximum and minimum values found.

Tests were performed at an average air velocity of 0.52 m/s in the channel. To optimize the separation of the particle sizes, the sample stream was offset from the centre of the channel by using unequal sheath air inlet pressures. The sheath air pressures were chosen such that the sample air would be centred about a position one-quarter of the channel width from the radially inner wall before the curved section of the main channel. After the traveling around the curved
section and into the region where the channel split, the smallest of the particles were evenly distributed between the radially inner and middle outlets, with 49% of these particles in the radially inner channel, 49% in the middle channel, and the remaining 2% in the radially outer channel. The higher diffusion coefficient of the 0.2 μm particles is a significant limitation to their separation into the radially inner channel.

The larger two particle sizes were both significantly separated into the middle and radially outer channels respectively, with 82% of the 1.0 μm particles were found in the centre channel and 87% of the 1.9 μm particles found in the radially outer channel (Figure 52).

6.5 Particle detector

The particle detector was tested with nebulized polystyrene latex microsphere; as particles collected on the filter between the photodiode and the light source, the photodiode output was recorded (Figure 54) and later compared to the particle concentration.

Figure 54. Sample PSAP data. Particles were 0.75 μm in diameter, with a pre-nebulized concentration of 0.01% w/v in water.

To allow for comparison between particles of different sizes, the particle area flux was found by multiplying the particle number concentration in air by the flow rate of the air and by the cross-sectional area per particle. The sensor response in ΔV/s was found by applying a linear regression to find the slope of the data. The sensor responded well over four orders of magnitude of particle concentration over time periods ranging from a few minutes for the high
concentrations to two hours for the low concentrations (Figure 55). A linear fit to the data in Figure 55 gave

\[ s = 1187 f - 1.15 \times 10^{-8} \]  

(33)

with \( s \) the sensor response in \( \Delta V/s \) and \( f \) the particle area flux in \( m^2/s \), with an \( R^2 \) value of 0.9.

6.5.1 Limit of detection estimates

The typical noise measured during the particle concentration tests was \( \pm 10 \) mV, as can be seen from the sample data in Figure 54.

The analog inputs on the DAQ used to measure the voltage signal has a maximum voltage of 10V with a 16 bit analog to digital converter (ADC); this means the resolution of the ADC is 0.15 mV, which means that the signal to noise ratio, rather than the resolution of the DAQ ADC, limits the resolution of the particle concentration that can be measured. The noise in the PSAP system was examined to ensure that there were no low-frequency noise effects which would affect a linear fit of the data. The fast Fourier transform of the sample data (Figure 56) shows that the only peak below 0.5 Hz is the near-zero frequency of the signal itself. This validates the use of a linear fit to quantify the data without any further signal processing.
The magnitude of the noise $n$ can be used to identify the lowest concentration detectable for a selected signal to noise ratio $R_{SN}$ and experiment run time $T_{test}$. For the linear regression which produces a curve fit with the form

$$a = m\gamma + b$$

with $a$ the dependent variable, $\gamma$ the independent variable, $b$ the intercept with the dependent variable axis, and $m$ the slope, the total signal change required

$$R_{SN}n = m\gamma T_{test}$$

(35)

can provide the minimum concentration

$$\gamma = \frac{R_{SN} n}{m T_{test}}$$

(36)

that can be detected, in units of m$^2$/s. If the flow rate $Q_{detector}$ is known, this concentration

$$\gamma' = \frac{R_{SN} n}{m T_{test} Q_{detector}}$$

(37)

can be converted to one of projected particle area per volume of air.

As an example, using a desired signal to noise ratio of 5, an average noise value of 10 mV, the value of $m$ found in the linear fit of equation 32, a one hour test time, and a flow rate of 1 L/min, the projected particle area per volume of air that can be detected is $7 \times 10^{-7}$ m$^2$/Lair. If the test aerosols were spherical 1 μm particles, this is a concentration of about $9 \times 10^5$ particles/L.
or 4.9 mg/m$^3$. This value is exceeds some of the exposure limits to respirable dust (see section 1.1.2). The detection limit could be decreased in several ways – the required signal to noise ratio could be lowered, the signal could be filtered to remove the noise, the test could be run for longer or at a higher flow rate, or the sensitivity of the device could be increased by decreasing the area of filter on which the particles land.

### 6.5.2 Error analysis for particle detection data

The vertical error bars in Figure 55 were obtained through error analysis of the particle detector data for each test. The error $\Delta f$ of the function $f = f(x_1, x_2, x_3, \ldots)$ is given by

$$(\Delta f)^2 = \sum_{i} \left( \frac{\partial f(x_1, x_2, x_3, \ldots)}{\partial x_i} \right)^2 (\Delta x_i)^2 \quad (38)$$

The number of particles per second, $N$, as a function of $M$ (change in mass in the liquid pre-aerosolized sample), $C$ (concentration, pre-nebulization in w/v), $F_{\text{corr}}$ (particle size-dependent correction factor from nebulizer efficiency tests), $d$ (particle diameter), $\rho$ (density), and $T$ (runtime) is

$$N = \frac{M}{4 \pi \left(\frac{d}{2}\right)^3 \rho} \frac{C F_{\text{corr}}}{T} = \frac{6CMF_{\text{corr}}}{\pi\rho d^3 T} \quad (39)$$

and so the error in $N$ is

$$(\Delta N)^2 = \left( \frac{\partial N}{\partial M} \right)^2 (\Delta M)^2 + \left( \frac{\partial N}{\partial d} \right)^2 (\Delta d)^2 + \left( \frac{\partial N}{\partial C} \right)^2 (\Delta C)^2 + \left( \frac{\partial N}{\partial F_{\text{corr}}} \right)^2 (\Delta F_{\text{corr}})^2 + \left( \frac{\partial N}{\partial T} \right)^2 (\Delta T)^2 \quad (40)$$

Similarly, the number of particles per second times the area, $N^*A$, is

$$N^*A = \frac{M\pi \left(\frac{d}{2}\right)^2}{4 \pi \left(\frac{d}{2}\right)^3 \rho} \frac{C F_{\text{corr}}}{T} = \frac{3CMF_{\text{corr}}}{2\rho dT} \quad (41)$$

so the error in $N^*A$ is
The partial derivatives of $N*A$ with respect to the variables $\Delta M$, $d$, $C$, and $F$ are:

$$\left(\Delta (NA)\right)^2 = \left(\frac{\partial N}{\partial M}\right)^2 (\Delta M)^2 + \left(\frac{\partial N}{\partial d}\right)^2 (\Delta d)^2 + \left(\frac{\partial N}{\partial C}\right)^2 (\Delta C)^2$$

$$+ \left(\frac{\partial N}{\partial F_{corr}}\right)^2 (\Delta F_{corr})^2 + \left(\frac{\partial N}{\partial T}\right)^2 (\Delta T)^2$$

(42)

The partial derivatives of $N*A$ with respect to the variables $\Delta M$, $d$, $C$, and $F$ are:

$$\frac{\partial N}{\partial M} = \frac{3CF_{corr}}{2\rho dT}$$

$$\frac{\partial N}{\partial d} = -\frac{3CMF_{corr}}{2\rho d^2 T}$$

$$\frac{\partial N}{\partial C} = \frac{3MF_{corr}}{2\rho dT}$$

$$\frac{\partial N}{\partial F_{corr}} = \frac{3CM}{2\rho dT}$$

$$\frac{\partial N}{\partial T} = -\frac{3CMF_{corr}}{2\rho d T^2}$$

The estimated errors for each individual variable are listed in Table 14.

Table 14. Error estimates for particle detector response variables.

<table>
<thead>
<tr>
<th>Variable</th>
<th>Estimated error [units]</th>
</tr>
</thead>
<tbody>
<tr>
<td>$\Delta M$</td>
<td>0.01 [g]</td>
</tr>
<tr>
<td>$\Delta d$</td>
<td>0.05 d [m]</td>
</tr>
<tr>
<td>$\Delta C$</td>
<td>0.01 C (w/v)</td>
</tr>
<tr>
<td>$\Delta F_{corr}$</td>
<td>estimated from the error of the correction factors</td>
</tr>
<tr>
<td>$\Delta T$</td>
<td>0</td>
</tr>
</tbody>
</table>

The error in mass was taken from the resolution of the scale used to measure the sample mass before and after, and from tests to determine how much sample would be left as a liquid on the nebulizer jet assembly when the system was turned off and the sample jar removed for weighing. The particle diameter error was from the manufacturer’s estimates, and the concentration estimate was also based on the resolution of the scale used to measure the sample.
while diluting. No time error was used, since the computer recording the data also takes time stamps at a resolution of hundredths of a second.

The largest error was found for the test performed at the lowest concentration. This concentration was in fact below the limit of detection found in the previous section’s analysis; it did not satisfy the signal to noise ratio imposed on that calculation.
7. DESIGN OPTIMIZATION

The simulations, initially written to match the experimental conditions, can also be broadened in their scope to inform future designs. By maintaining the general approach of the system but varying the channel geometry and operating parameters, a general set of relationships between the design and operating variable and the resulting implications can be explored. These can be used to select between the various tradeoffs present in the system and to optimize the design for a particular range of particle sizes. An overview of the design and operating variables and the various outcomes resulting in changes in those variables is given in Table 15.

<table>
<thead>
<tr>
<th>Design &amp; operation variables</th>
<th>Results / implications to consider</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Design variables:</strong></td>
<td>Separation characteristics</td>
</tr>
<tr>
<td>• channel width</td>
<td>• range of particles sizes with best separation resolution</td>
</tr>
<tr>
<td>• channel height</td>
<td>• maximum particle size that can be used</td>
</tr>
<tr>
<td>• radius of curvature of bend</td>
<td></td>
</tr>
<tr>
<td>• angle of bend</td>
<td></td>
</tr>
<tr>
<td><strong>Operating variables:</strong></td>
<td>Flow characteristics</td>
</tr>
<tr>
<td>• mean velocity</td>
<td>• total throughput of sample air</td>
</tr>
<tr>
<td>• ratio of sheath air to sample air flow rates</td>
<td>• development of secondary Dean flow</td>
</tr>
<tr>
<td>• particle sizes</td>
<td>• development of slip flow at walls</td>
</tr>
<tr>
<td></td>
<td>System implications</td>
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<td></td>
<td>• precision in inlet pressure control</td>
</tr>
<tr>
<td></td>
<td>• overall size of device</td>
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<td>• ease of fabrication</td>
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Because of the complex interplay of variables, two different overall sets of simulations are presented here. The first set of simulations holds constant the curved channel geometry as it is was used in the experiments in this work. The velocity is varied and the separation resolution, particle size limitations, and losses due to centrifugal forces and diffusion are simulated and presented. The second approach is to vary the width of the separation channel, and explore the effect this has on each of the outcomes listed above at a few different velocities. After these results are shown, a brief discussion of the effects of the other variables and outcomes mentioned in Table 15 is presented.

7.1 System characteristics with constant channel geometry

The constant geometry simulations use the same channel geometry as the experiments and earlier simulations presented in this work. The channel height is 100 μm, the width and
length of the main separation section are 60 μm and 4 cm, respectively, and the curved section of the channel has a 180° bend with a radius of curvature of 2.5 mm. Figure 57 and Figure 58 show the results of simulations providing the mean position of particles in the width of the channel, as a fraction of the total channel width. The ratio of sheath to sample air for these and the rest of the simulations in this chapter is 8:1. In Figure 57, a wide range of average air velocities is used: the velocities range from 0.25 m/s to 6 m/s and the particle diameters from 0.25 μm to 2 μm. Figure 58 focuses on the velocities in the lower range of those in Figure 57, and includes particles up to 3 μm in diameter. All particles are assumed to have the same physical properties as polystyrene latex microspheres.

![Graph](image)

**Figure 57.** Simulated mean position of particles in the width dimension of the channel, as a fraction of the total channel width, for average air velocities ranging from 0.25 m/s to 6 m/s and particle diameters from 0.25 μm to 2 μm.
In Figure 57, it is clear that particles larger than about 1.5 µm achieve their position of maximum possible separation by an average channel air velocity of about 1 m/s. Beyond this velocity, these particles quickly become indistinguishable from any larger particles as they all cluster towards the radially outer wall of the channel. In reality, many of these particles will end up depositing on the radially outer wall, which will then effectively decrease the channel width and cause other radially-outward particles to accumulate, causing a cascading failure resulting in complete channel blockage. This deposition of particles on the channel walls was shown in simulations of the actual particle sizes used experimentally in section 3.3.1, and will be further explored later in this section. The data in Figure 58 reveal more closely the separation effects in this 0-1 m/s average air velocity range. Over this range, the final mean particle position in the channel appears nearly linear for sub-micron particles, as they are only moved across a distance less than one third of the channel width. The particles over 1 µm in diameter can be handled and

Figure 58. Simulated mean position of particles in the width dimension of the channel, as a fraction of the total channel width, for average air velocities ranging from 0.1 m/s to 1 m/s and particle diameters from 0.25 µm to 3 µm.
separated at velocities of less than about 0.4 m/s in general, but their final position varies with small changes in average air velocity.

The particle diameters used to generate these plots were evenly spaced apart from one another. The results of these simulations can be used to identify the particle diameter which, at a given velocity, represents the diameter of highest separation resolution. This was accomplished by identifying for each velocity the particle size that occupies a final mean position the farthest on average from its two neighbouring particle sizes in the set. The results of this are shown in Figure 59 and Figure 60 for the 0-6 m/s and 0-1 m/s average air velocity sets, respectively. The data appear somewhat discontinuous but this is merely a result of the 0.25 µm difference in particle sizes used in the simulations. In both datasets, the particles on the high end of the range of simulated particle sizes are the best separated at the lowest velocities. These plots can be used if particles of a certain range are to be separated in the channels, to determine what velocity or range of velocities would be most appropriate for testing.

![Figure 59](image-url)  
**Figure 59.** Particle diameter with the highest separation from particles of its diameter ±0.25 µm in the 0-6 m/s average air velocity range, based on the data in Figure 57.
The velocity can be chosen to optimize separation resolution for a certain particle size, but another consideration that must be taken into account is the deposition of particles onto the channel walls described above. Particles that are too large for a certain air velocity and channel geometry will impact against the radially outer side wall, and small particles can diffuse onto the channel walls – particularly the top and bottom walls, as the small particles are less likely to come near a side wall at the lower velocities where they diffuse more. The deposition of particles on the channel walls was simulated for velocities from 0.1 m/s to 1 m/s in 0.1 m/s intervals and with particle of diameters 0.25 µm – 3 µm in 0.25 µm interval (Figure 61). It was assumed that if the centre of any particle came to within a distance to the wall of its own radius, it would deposit on the wall. At the lowest velocity simulated (0.25 m/s), up to 30% of the simulated particles deposited on the channel walls by diffusion, with the effect the greatest for the smallest particles and decreasing with increasing particle size. For each velocity above 0.2 m/s, it is clear that there is a critical diameter at which particles will rapidly accumulate on the radially outer sidewalls due to centrifugal forces. This provides an estimate of the upper limit of the particle diameter that can be used in the channel with a given average air velocity.

Figure 60. Particle diameter with the highest separation from particles of its diameter ±0.25 µm in the 0-1 m/s average air velocity range, based on the data in Figure 59.
Since particle deposition by diffusion may also be undesirable for some applications, summaries are presented in Figure 62 of the range of diameters of simulated particles that were successfully transported through the simulated channel with one of two criteria for successful transportation. The first criteria (Figure 62a) requires that fewer than 10% of the particles were deposited on the walls by diffusion or centrifugal force. The second criteria (Figure 62b) stipulates that 15% of the particles may be deposited on the top, bottom, and inner walls but that fewer than 5% of the particles may be deposited on the outer wall. This second criteria was chosen because the deposition of particles on the outer wall by centrifugal force can quickly cause a total channel failure, while the deposition on the other three walls by diffusion, more
spread throughout the length of the channel, may still allow the channel to function despite higher rates of particle deposition.

7.2 System characteristics with varying channel width

Each of the effects simulated in the previous section were explored but with a varying channel width. The diameter of particles with the highest separation from those with diameters ±0.25 µm is presented as a function of channel width (30 µm, 60 µm, 100 µm, or 150 µm) and average velocity (0.2 m/s to 1.1 m/s in 0.3 m/s intervals) in Figure 63. This diameter of

![Graph](attachment:image.png)

Figure 62. Smallest and largest diameter of simulated particles that were transported through the channel with (a) fewer than 10% of the particles deposited on the walls by diffusion or centrifugal force and (b) fewer than 15% of the particles deposited on the top, bottom, and inner walls and fewer than 5% of the particle deposited on the outer wall.
maximum separation resolution increases with increasing channel width and as the average air velocity in the channel decreases. The five-fold increase in channel width across the data presented results in an approximate doubling of this optimal particle diameter at the lowest velocity (0.2 m/s), and results in a four-fold increase in this particle diameter at the highest velocity (1.1 m/s).

The simulations of particle deposition on the channel walls were also repeated with the channel width as a variable. Figure 64 shows the maximum particle diameter that can be run through channels of widths 30 μm, 60 μm, 100 μm, and 150 μm at velocities of 0.2 m/s to 1.1 m/s in 0.3 m/s intervals. The maximum diameter that was successfully transported through the channel depended strongly on both the channel width and the average air velocity. Increasing the channel width by five times resulted in a three-fold increase in the maximum allowable particle diameter.

Figure 63. Particle diameter with the highest separation from particles of its diameter ±0.25 μm as a function of the channel width (30 μm, 60 μm, 100 μm, or 150 μm) and average velocity (0.2 m/s to 1.1 m/s in 0.3 m/s intervals).
The data of Figure 63 and Figure 64 are combined in Figure 65 to show the maximum and optimal particle diameter for the four velocities (0.2 m/s, 0.5 m/s, 0.8 m/s, and 1.1 m/s) and four channel widths (30 μm, 60 μm, 100 μm, and 150 μm) used in the simulations. The simulated particles had diameters that were multiples of 0.25 μm in diameter, so the data points are approximations to the values that could be found with a finer resolution in diameter. Overall, the distance between the maximum and optimal separation diameter is larger for wider channels, and for lower velocities.

Figure 64. Largest diameter of simulated particles that were transported through the channel with fewer than 5% of the particles deposited on the radially outer wall.
Other causes and effects

The design variables, flow characteristics, and system implications not explored in the previous two sections are discussed here in the context of their significance in system optimization. The effect of the channel’s height, aspect ratio, radius of curvature, and angle of bend are covered.

Figure 65. Largest allowable diameter particles and diameter of maximum separation resolution, for various channel widths at velocities (a) 0.2 m/s, (b) 0.5 m/s, (c) 0.8 m/s, (d) 1.1 m/s

7.3 Other causes and effects

The design variables, flow characteristics, and system implications not explored in the previous two sections are discussed here in the context of their significance in system optimization. The effect of the channel’s height, aspect ratio, radius of curvature, and angle of bend are covered.
7.3.1 Channel height

The channel height can be varied quite a bit in theory, but with less benefit to separation than the width variations. A decrease in channel height means a higher-resistance channel, with typically lower flow rates and a closer proximity of particle to the top and bottom surface of the channel. Increasing the height would decrease the flow resistance, resulting in a higher flow rate and velocity for a given set of inlet pressures; this would require higher precision control without offering significant benefits. Additionally, any height change needs to be considered in terms of the channel aspect ratio and hydraulic diameter. Fabricating microchannel molds with aspect ratios of higher than about 5:1 (height:width) would not be feasible with SU-8 3050 photoresist; an alternative photoresist or fabrication process would be required. To avoid the occurrence of slip flow, the Knudsen number (section 2.4.2) should remain below a threshold value. Various values have been proposed; a conservative selection is Kn < 0.001. This requires that the hydraulic diameter of the channel remain larger than 67.5 μm.

The equivalent minimum height

\[ H = \frac{1}{2} \left( \frac{1}{67.5} - \frac{1}{2W} \right)^{-1} \]  

(43)

to satisfy this criteria for varying channel widths W, with both in units of micrometers, are shown in Figure 66. This criteria is not achievable at aspect ratios of less than 5:1 for width of less than 45 μm.

![Figure 66. Minimum channel height to satisfy Kn<0.001 for various channel widths.](image-url)
7.3.2 Channel bend radius

The radius of curvature $r$ can both be modified to change the device performance; the radial velocity $U$ of a particle depends inversely on $r$. If particle test size and the other channel geometry factors are kept constant, a smaller radius of curvature requires the use of lower velocities to prevent particles from impacting on the radially outer sidewalls. A larger radius of curvature has the opposite effect, but also means that the device’s physical footprint would be larger. The main limitation on the radius of curvature is the possibility of developing secondary Dean flows in the channel (section 2.4.1). The Dean number is inversely proportional to the root of the radius of curvature. The fluid velocity and the channel width, height, and radius of curvature all factor into the Dean number as well. If, for example, the radius of curvature was decreased, the only way to prevent the Dean number from increasing would be to decrease the channel width and increase the height. This would then decrease the maximum size of particle testable in the channel at a given flow rate. In summary, a decrease in bend radius has the advantage of a smaller device footprint, but at the cost of either decreasing the range of particles sizes that can be used or increasing the required accuracy of the inlet pressure control.

7.4 Three-dimensional flow focusing

Focusing the flow in the height direction as well as in the width direction would improve the separation resolution significantly, by reducing the probability of particles residing in the low flow regions near the top and bottom wall of the channel. Particles in these regions are not well-separated because of the decreased centrifugal force resulting from the decreased axial velocity; this phenomenon can be seen, for example, in Figure 12.

Figure 67 shows a simulation of particles of diameter 0.25 $\mu$m, 0.5 $\mu$m, 0.75 $\mu$m, and 1 $\mu$m traveling in a channel with identical geometry to that tested experimentally in this work, with two- and three-dimensional focusing. For the two-dimensional focusing case, the ratio of flow rates of sheath to sample air was 8:1, resulting in a starting width of the particles of 7% of the full channel width. For the three-dimensional focusing case the same width was used, and the region defining the starting particles of the particles was confined to 7% of the full channel height, centered in the height dimension of the channel. The separation of the particles is significantly improved with the three-dimensional focusing, but the realization of this approach requires a much more complicated microchannel fabrication process.
Figure 67. Separation of 0.25 μm, 0.5 μm, 0.75 μm, and 1 μm particles at an average flow velocity of 1 m/s with (a) two-dimensional flow focusing, as was used in this work, and (b) three-dimensional flow focusing, which improves the separation resolution of the system. For (a), the ratio of flow rates of sheath to sample air was 8:1, resulting in a starting width of the particles of 7% of the full channel width. For (b), the width and height of the region defining the particle starting positions was 7% of the full width and full height of the channel respectively.
8. CONCLUSIONS AND FUTURE WORK

8.1 Conclusions

8.1.1 Particle separation

A microchannel-based system for the size-based separation of aerosols has been presented. The curved channel design imposes centrifugal forces on the aerosols, which travel radially outwards at a velocity dependent on their diameter and mass. The system was tested with fluorescent airborne polystyrene latex microspheres, for which the mass varies only with the diameter. The test aerosols were collected on filters placed in the path of the particles at the channel outlet; their fluorescence properties allowed them to be imaged and distinguished by size with an epifluorescence microscope. Straight channels were fabricated to confirm the repeatable functionality of the experimental setup, to validate the model of flow focusing, and to compare the measured effects of diffusion to the simulated effects. Curved channels with a single outlet were used to test particles in the range of 0.2 µm to 1.9 µm at a variety of velocities. They were also used to test the separation of a mixture of 1.9 µm particles and 3.2 µm particles, to determine the device’s suitability for PM2.5 measurements. Curved channel devices with two and three outlets were tested for their ability to separate particles by size into two or three different sub-outlets. A mixture of 0.2 µm particles and 0.75 µm particles were used in a two-outlet channel, with 86% of the larger particles traveling to the radially outer channel. A mixture of three particle sizes, 0.2 µm, 1.0 µm, and 1.9 µm, were used to test the three-outlet device; the larger two particle sizes were both significantly separated into the middle and radially outer channels respectively, with 82% of the 1.0 µm particles were found in the centre channel and 87% of the 1.9 µm particles found in the radially outer channel. In both the two and three outlet devices, the smallest (0.2 µm) particles were present in multiple outlet channels, as the ability to separate those particles is limited by their high diffusion coefficient.

In general, the experiments showed good correlation with the simulated results. The number of particles handled in each experiment was typically fewer than was used in the simulations, resulting in greater variation in the results and accounting for some of the difference between the simulations and experiments.

When optimizing the operating parameters for a certain range of particle sizes, several factors must be balanced for successful separation. Holding the geometry of the device constant, increasing the flow rate results in having a larger total volume of air sampled, which is a
desirable result for aerosol monitoring. The increased inlet pressure for this case also means less fine control is required, which allows for the use of simpler and cheaper control system. The particle diameters with the highest separation resolution shifts towards the smaller particles as the velocity increases, but the maximum particle size that can be run through the channel decreases.

The no-slip condition, resulting in low velocities near the wall, causes both benefits and detriments to the separation process. The low velocity near the side walls helps to prevent the impaction of particles against the walls, since the centrifugal force pushing the particles outwards is depending on the axial velocity. However, the low velocity at the top and bottom of the channel prevent particles in those positions from being size-sorted.

A more fundamental limitation to this size separation system, one which must be addressed with any particle separation or detection system, is that real aerosols are not all perfect spheres of equal density. The choice of detection system could help to provide some more information about the shapes and densities of the particulates being measured by the system, and could possibly be used to correct the measurements for these variations.

This approach is the first time that aerosols have been separated by centrifugal forces in a microchannel, and one of very few approaches that have been used for any kind of size-based separation of airborne particles in microchannels. The small footprint and potential for integration offered by microsystem fabrication technology make it a desirable avenue of pursuit for the development of small, portable particulate monitors. The results presented here confirm that this approach to size-separation is a feasible option for a future microsystem-based size-selective particulate monitor.

8.1.2 Detection system

The approach to particle detection presented here offers both advantages and disadvantages as a candidate for integration with the particle separation system. The primary advantage lies in its simplicity. The light source – filter – light detector arrangement is a straightforward one, and can be easily incorporated into the existing PDMS microchannels by embedding the light source into the PDMS before curing and by using the filters already collecting particles to do so for detection. In addition, the cumulative nature of the filter collection system decreases the need for high-resolution low-noise optical systems, but with the disadvantage of eventually saturating the filter. The other chief disadvantage is one of scaling - the testing of the particle detector has been at flow rate levels far higher than are reasonable for
use in the microchannel separator. To determine the suitability of the detection system from a scaling perspective, the relative flow rates and active filter areas can be compared.

The flow rate through the particle detection system was on the order of 0.5 L/min. The flow rate of particle-laden sample air through the microchannel was typically around 2 µL/s. The resulting ratio of flow rates was around 4000:1. The diameter of the region of filter exposed to the test air in the detector was 5 mm; this is equal to an area of \(2 \times 10^{-5} \text{ m}^2\). If a circular region of filter is also used to collect particles in the microchannel, it would need to have an diameter of no more than 40 µm to maintain the same ratio of flow rates to particle collection surface area. An even smaller area would improve the limit of detection further. For this estimate to hold value, it is necessary to assume that an alternative light source can be found, which can provide the same total light energy over a much smaller region than did the original detection system LED. A laser diode might be suitable for the purpose, as the beam divergence angle of a laser diode is typically lower than that of an LED.

### 8.2 Future work

#### 8.2.1 Improving device performance

The device performance would be significantly improved by incorporating height-wise flow focusing along with the width-wise focusing. The low velocity at the top and bottom of the channel prevent particles in those regions from being size-sorted and also result in regions with higher diffusion, but having particles travel in these regions is inevitable with two-dimensional focusing. With three-dimensional focusing, as exemplified in section 7.4, the particles could be confined to a central region of the channel in both the width and height directions, thereby significantly improving the resolution limitations of this separation technique.

The material selection for the device would also be a consideration, both for integration purposes and for inherent performance improvements. Specifically, PDMS is an electrical insulator and therefore prone to attracting particles due to electrostatic effects, decreasing the transmission efficiency of the channel. Using a conductive material would decrease these electrostatic losses. This could be achieved in one of several ways: a moldable conductive polymer could replace the PDMS used here, PDMS could be used but coated or combined with a conductive material, or an entirely different fabrication technique could be used, in which a curable polymer was not required.
8.2.2 Integration and scalability

The ideal future system incorporating this separation approach would benefit from few additional features. First, even with the highest flow rates used in this work, the air sampling rate remains a significant limitation of the device. Monitoring applications may require the use of several separation channels in parallel to achieve higher overall flow rates; the channel dimensions could also be increased if sufficiently accurate inlet pressure control could be provided. It would also be necessary to have some sort of preliminary system removing particles of sizes large enough to block the channels; a micromachined virtual impactor upstream of the channel would be one option, as it could remove the overly large particles through a minor flow and allow only particles smaller than a maximum size cutoff to pass on to the separation channels for higher resolution size separation. Running multiple systems in parallel with preselectors of different cutoff sizes could allow some of the tradeoffs of particle size and separation resolution to be overcome. Some separation channels could have preselectors with smaller cutoff sizes and lower resistance inlets, so that smaller-sized particles experience separation at high velocities. Other channels could accept larger particles and separate those at lower velocities, without having the lower resolution of the smaller particles under those circumstances be an issue. By using the inlet channel geometries to control the relative inlet flow resistances and therefore the flow rates, a single pressure controller could be used for multiple channels with varying mean velocities.

Other options for particle detection could also be explored, including a single-particle detection system with integrated photodetectors of sufficient sensitivity to measure the scattering of light from a single particle, or mass-based sensors which could provide information the mass concentration of the particles traveling in each size bin.
BIBLIOGRAPHY


[22] U.S. Occupational Safety and Health Administration, "Occupational Safety and Health Standard # 1910.1000 TABLE Z-1 Limits for Air Contaminants".


APPENDIX 1: SIMULATION BOUNDARY CONDITIONS

The choice of boundary conditions affects the final shape of the particle simulations, and this effect is particularly noticeable for the curved channel simulations. In the simulations shown in Figure 68, the particles were removed from the simulation if they came to within a particular distance of the channel walls. This distance was varied from 1 to 10 μm, to visualize the effect this has on the shape of the particle distribution. The condition had no notable effect on the position of the peak particle concentration, but, unsurprisingly, had a direct influence on the left edge of the distribution, representing particles located near the top and bottom of the channel.

![Graph showing particle concentration across channel width.](image)

Figure 68. Simulation results for 1.9 μm particles in a curved microchannel, with an average axial air velocity of 0.5 m/s. The particles were removed from the simulation if they came to within a certain distance of the wall, with this distance as indicated by the plot legend.

If the boundary conditions are instead changed so that particles which approach the wall bounce back to a position a few microns from the wall, this creates an artificial peak of particles in these near-wall regions, seen near the centre of the channel width. The difference between these two approaches is shown in Figure 69.
Figure 69. Simulation of 0.75 μm and 1.9 μm particles traveling in a channel with an average air velocity of 0.5 m/s. In (a), particles which come to within half of their own diameter of the wall are forced back towards the channel centre to a distance just larger than their own radius. In (b), particles which come to within half of their own diameter of the wall are removed from the simulation.
APPENDIX 2: SAMPLE PARTICLE DISTRIBUTION DATA

Sample data from the diffusion experiments (see setup, section 5.4, and results, section 6.3) are presented here. The data shown is 0.2 µm and 1.0 µm particles at the lowest and highest velocities tested, as examples of the extremes of the range over which the tests were performed.

0.2 µm particles, 0.7 m/s average velocity

![Graph](image1)

0.2 µm particles, 1.4 m/s average velocity

![Graph](image2)
1 μm particles, 0.7 m/s average velocity

1 μm particles, 1.4 m/s average velocity