SANDIA REPORT

SAND2001-8381
Unlimited Release
Printed June 2001

Miniature Bioaerosal Concentrator: Progress Through FY2000

Karl Wally

Prepared by
Sandia National Laboratories
Albuquerque, New Mexico 87185 and Livermore, California 94550

Sandia is a multiprogram laboratory operated by Sandia Corporation, a Lockheed Martin Company, for the United States Department of Energy under Contract DE-AC04-94AL85000.

Issued by Sandia National Laboratories, operated for the United States Department of Energy by Sandia Corporation.

Available to DOE and DOE contractors from
U.S. Department of Energy
Office of Scientific and Technical Information
P.O. Box 62
Oak Ridge, TN 37831

Telephone: (865)576-8401
Facsimile: (865)576-5728
E-Mail: reports@adonis.osti.gov
Online ordering: http://www.doe.gov/bridge

Available to the public from
U.S. Department of Commerce
National Technical Information Service
5285 Port Royal Rd
Springfield, VA 22161

Telephone: (800)553-6847
Facsimile: (703)605-6900
E-Mail: orders@ntis.fedworld.gov
Online order: http://www.ntis.gov/ordering.htm

SAND2001-8381
Unlimited Release
Printed June 2001

Miniature Bioaerosol Concentrator: Progress through FY2000

Karl Wally
Exploratory Systems Technologies Department

Sandia National Laboratories
P.O. Box 969
Livermore, California 94551-0969

Abstract

This report describes progress made on the design and development of a miniature bioaerosol concentrator intended to concentrate the respirable fraction of a small bioaerosol sample directly into a microliter liquid volume contained in open micro-channels within the device. The chosen design concept is described, technical issues related to its development are discussed, and research and development progress through FY2000 is reported. This work was funded under Sandia's Laboratory Directed Research and Development (LDRD) program as Project 10371. This report has been prepared in fulfillment of Department of Energy reporting requirements for the LDRD program.

Acknowledgements

The author would like to gratefully acknowledge the substantial contributions provided by Daniel J. Rader of Thermal/Fluid Experimental Sciences Department 9112. Dan served as co-principal investigator in the area of aerosol science; he has provided original concept descriptions, performed FIDAP simulation calculations, conducted aerosol experiments, and prepared original figures for the Virtual Cyclone, Opposed Flow Virtual Cyclone, and the Focusing 2-Dimensional Wedge Impactor.

Contents

List of Figures

List of Tables

Miniature Bioaerosol Concentrator: Progress through FY2000

Background

Events over the last decade have highlighted a need for the United States to be better prepared to deal with the potential threat of chemical and biological warfare agent aerosols. Concerns arising during the Gulf War, and later, surrounding the Japanese terrorist cult Aum Shinri Kyo, have highlighted a need for better systems capable of detecting the presence of chemical and biological warfare agent aerosols. Toward that end, the DOE and the national laboratories have undertaken the research and development of technologies for providing improved detection capabilities. To be most useful, emphasis has been placed on hand-portable detector systems.

Biological warfare agent (BWA) aerosols present an especially insidious threat. Such biological aerosols may contain pathogenic bacteria or spores, or may consist of particles of an inert matrix such as clay infused with biotoxins which have been derived from biological systems. Biological warfare agent (BWA) aerosols can be a deadly threat even at concentrations as diffuse as one pathogenic particle per liter of air. To be effective, therefore, BWA sensors must be capable of detecting BWA aerosols even at these extremely low concentrations. The most capable portable real-time BWA detectors under current development employ techniques of direct analytical detection by biological or chemical means in aqueous or solvent solutions. They are often variants of capillary-based chromatography techniques miniaturized to the microfluidic level. These techniques may require a minimum sample concentration of ten bio-particles per microliter of sample solution to achieve a minimal detection threshold. Sample concentrations more dilute than this simply risk being too dilute for reliable detection.* Worse, the entire volume of a single sampling to be analyzed by the microfluidic detection system may routinely consist of 1 microliter of fluid or less. There is, consequently, a considerable need for a sampling device that can serve as a "front end" to such hand portable detectors; a device that can help to reconcile the conflicting conditions which arise in the operation of such microfluidic detectors. That is, there is a need for a device to collect highly diffuse bioaerosols and concentrate them into microfluid sample volumes to a threshold concentration level sufficient for further analysis and detection.

Furthermore, for use in hand-portable detectors, such a sampling device must be both battery operable and itself miniature – only a few cubic inches in volume. Also, because many detection techniques employ bioaffinity principles that involve chemical and biochemical interactions with surface proteins of the collected particles, it is desirable

*This immediate discussion applies to direct analytical chemical detection or biological assay techniques; it does not address the requirements or capabilities of DNA techniques such as polymerase chain reaction.

that the sampling device employ only benign collection techniques that avoid damage to these surface proteins (damage as might occur, for example, in corona discharge in an electrostatic precipitator).

Additionally, because the need is to concentrate aerosol particles into such tiny fluidic volumes, it may be preferable not to collect the entire content of the environmental particle burden routinely encountered. Rather, it might be preferable that the sampling device be able to screen out particles unlikely to be BWA aerosols, and to concentrate only that fraction most likely to be BWA aerosols. Collecting only *respirable* particles can do this. *Respirable* particles are those that are less about 5 microns in size, but greater than about 1 micron in size. Particles larger than 5 microns tend to be less persistent as aerosols, therefore settle-out rapidly. Also, they are more likely to be filtered by the body's natural filtering mechanisms (nose hairs, mucous, sneeze reflex, etc.). Particles smaller than 1 micron tend to be exhaled as readily as they are inhaled. For these reasons, BWA aerosols have historically been engineered to have particle sizes that fall within the respirable range, to produce a maximized likelihood of lodging deep within a victim's lungs. Coincidentally, many BWA bacterial cells and spores fall naturally within this range.

Miniature Bioaerosol Concentrator Design Concept

From all these considerations, it may be observed that satisfying the requirements of battery-powered hand-portable microanalysis systems imposes staggeringly challenging specifications upon a miniature bioaerosol concentrator. To perform prompt measurements, a concentrator must be capable of sampling a representative bioaerosol (i.e., one BWA particle per liter of air) at a rate of about 10 to 20 liters of air per minute, with a high collection efficiency for respirable particles (1-5 microns diameter).

These particles must then be concentrated into a mere one microliter (or less) of aqueous solution. Starting with a one microliter sample, micro systems which require even smaller samples (e.g. tens of nanoliters) can at least contemplate incorporation of a further concentration step (e.g. via micro-evaporation or micro-filtration) that may prove workable. A significantly larger sample than one microliter renders all such micro-concentration approaches futile. This step of concentration of sample particles into microliter fluid volumes is perhaps the single most dauntingly difficult proposition within the proposed concentrator concept. No current devices, either commercially available devices or developmental R&D devices, even approach this level of performance.

Too, the concentrator should consume as little power as possible. A target power consumption of 1 watt or less represents a challenging target compatible with battery powered hand-portable detection systems.

Finally, aside from these physical performance requirements, the detector should use benign collection techniques capable of leaving the particles and their surface proteins undamaged by collection.

To satisfy these requirements, we proposed the miniature bioaerosol concentrator concept shown in Figure 1. The envisioned device is comprised of a microfabricated aerosol concentrator sandwich through which sample air is drawn by means of an integral vacuum fan and brushless d.c. pancake motor. The microfabricated aerosol concentrator sandwich is itself comprised of two microfabricated parts, a micro-separator to aerodynamically separate particles from the bulk air flow, followed by a micro-impinger in which the separated particles are impinged into collection fluid in open micro-channels. We proposed to use electro-osmotic flow principles to move the collection fluid through the open micro-channels.

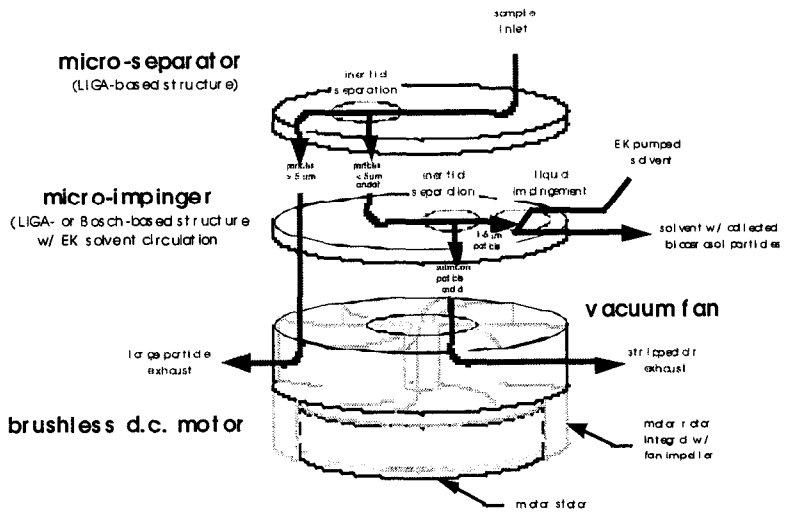

Figure 1. Miniature Bioaerosol Concentrator Concept Design

The concept incorporates several highly challenging elements of performance. The micro-separator must not only separate aerosol particles in the respirable size range, but also concentrate and focus those particles through the tiny aperture of the microfluidic channels of the micro-impinger. The micro-impinger must provide adequate fluid to collect the particles and compensate for evaporation losses, yet minimize the overall fluid sample volume to be compatible with sample volume needs of microfluidic analysis systems. Finally, the air flow passage through the system must be sufficiently open to achieve the desired flow rate with a low powered miniature fan and motor. Each of these requirements separately challenges the state-of-the-art. Their integration into a single miniature device, therefore, represents a high-risk proposition suitable for pursuit under the Laboratory Directed Research and Development (LDRD) program.

Progress through FY2000

Synopsis

Work began on this project in FY1998 with LDRD funding support from the LDRD project selection committee of Sandia's Nonproliferation Strategic Business Unit (i.e., the (NP&MC SBU*). Due to the risk inherent in the original proposal, the NP&MC SBU LDRD project selection committee provided only partial funding for development in the first year, asking that those limited funds be directed at better establishing the feasibility of the proposed concept. Feasibility was established to the committee's satisfaction by exploring the proposed performance on two fronts. First, we employed numerical modeling of the proposed particle separation scheme to predict that we could indeed achieve the desired particle separation, concentration, and focusing performance. Second, we demonstrated the ability to establish suitable fluid flow in closed micro-channels via electro-osmotic flow principles. The project was subsequently funded as a continuing LDRD project for FY1999 and FY2000. Work in FY1999 concentrated primarily on independent development of the micro-separator and micro-impinger portions of the miniature device. In FY2000, in addition to continuing development of these key components, the two separately developed elements where brought together to begin development of an integrated device.

As of the end of FY2000, we have successfully demonstrated many key elements of the miniature bioaerosol concentrator concept, but we have not yet been able to demonstrate successful operation of a micro-impinger exhibiting the desired behavior, and therefore have also not yet demonstrated operation of an integrated concentrator device. This lack of success with the micro-impinger has been largely the result of problems maintaining continuous fluid priming in open micro-channels. The problem is one of an imbalance between evaporation and capillary wicking; the imbalance is due primarily to a compromise in micro-channel aspect ratio. The inability to achieve consistent electro-osmotic function in a truly photo-etchable glass meant that we had to abandon the tall aspect ratio we desired, for the wide aspect ratio that is unavoidable in conventional etching of glass. A number of potential work-arounds to this problem have been under investigation. The problem and solutions under investigation are described in greater detail later.

Our best success on the miniature bioaerosol concentrator project to date has centered on the micro-separator portion of the concentrator, dealing with particle separation, concentration, and focusing. Several successful geometries have been investigated and demonstrated.

NP&MC SBU funding for this project will not be continued for FY2001, in keeping with an established LDRD policy of not pursuing projects beyond a total of three years of

* Nuclear Proliferation & Materials Control. The purview of this strategic business unit includes the nonproliferation of all forms of weapons of mass destruction, including chemical and biological weapons.

funding. Nevertheless, work on the project has been able to continue into FY2001 with modest funding from an alternate project source. That project is hoping to incorporate a successful miniature bioaerosol concentrator into yet another type of analysis and detection technology. In FY2001, we are continuing our pursuit of solutions to the micro-impinger problems.

Micro-Separator Development

The purpose of the micro-separator is to separate bioaerosols in the 1 - 5 micron diameter range, so-called respirable particles, from larger and smaller particles, and to concentrate and focus these respirable particles for subsequent deposition on or in some sort of collector or impinger. We find the most promising approach is to use inertial forces to separate and concentrate particles based on their aerodynamic size. One such device that uses inertial principles for aerosol separation and concentration is the *aerodynamic lens*. It concentrates particles along the centerline of a flow channel by a series of flow contractions and expansions. If the particles' aerodynamic sizes are less than a critical size, each contraction steers particles toward the channel centerline. Particles larger than the critical size move farther from the centerline. Another scheme is the *virtual impactor*, which generally consists of a jet impinging on a plate normal to the jet, which plate has a small hole located at the jet centerline. If flow through the hole is restricted, the hole acts as a virtual surface, so that the impinging jet turns and flows outward radially. Because of their inertia, particles cannot make the turn and are impacted into the virtual surface, concentrating the particles into the space below. However, both the aerodynamic lens and the virtual impactor have drawbacks that make them less than ideal for miniature or micro-miniature application. These drawbacks include wall losses and complexity of design.

In 1997, Sandia developed the *virtual cyclone* for particle concentration via inertial separation [1]. The virtual cyclone separates particles from a main flow and concentrates them in an adjacent chamber where they recirculate. In the virtual cyclone, the main particle-laden flow follows a wall that curves away from the original flow direction, as shown in Figure 2. A wall forms an inward-curving boundary to the main flow, while the outer boundary is formed by an adjacent confined recirculating flow into which particles move by centrifugal action. Thus, in the virtual cyclone, particles are separated from the main flow by crossing a dividing streamline that separates the main flow from the adjacent recirculating flow. An especial merit of the virtual cyclone is that it greatly reduces particle deposition on walls. The major drawback is that no convenient method presents itself for removing particles from the recirculating flow.

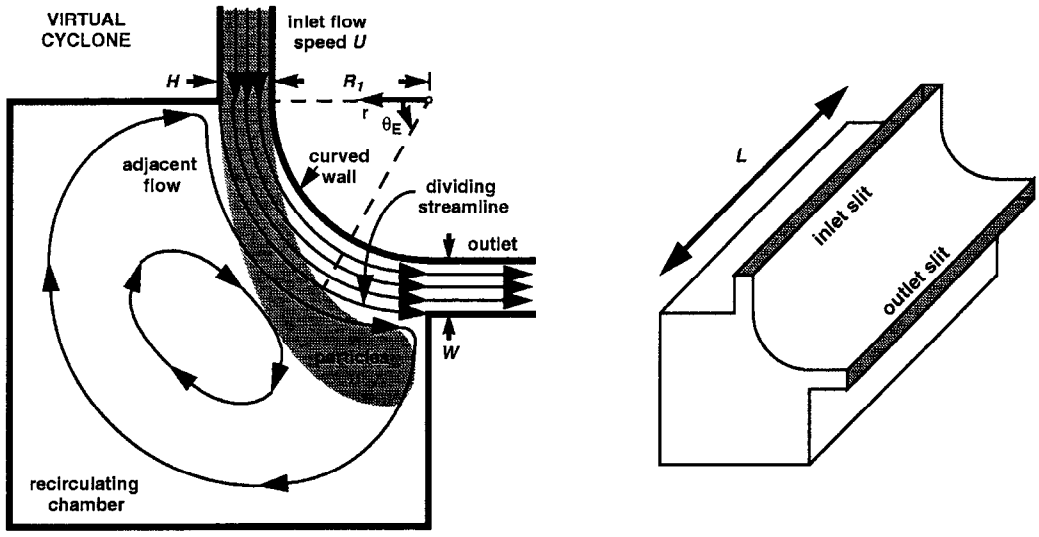

Figure 2. **Schematic diagrams of the standard virtual cyclone: a) side view, b) perspective view.**

Opposed–Flow Virtual Cyclone

Under this LDRD, we have developed a new method for particle concentration by inertial separation, which we have called the *opposed-flow virtual cyclone* (OFVC). The OFVC is a variation on the virtual cyclone theme. It preserves inherent advantages (no-impact particle separation in a simple geometry), while providing a more convenient design for concentrating particles in a flow-through geometry. Numerical fluid/particle simulations (using the commercial flow solver FIDAP, published by Fluent International) of several 2-Dimensional OFVC designs have been completed. The simulations show that 2-D OFVC devices should provide significant concentration along the plane of symmetry for particles of 1 to 5 micron diameter. A schematic of one embodiment of an OFVC is shown in Figure 3. The device consists of two virtual cyclones arranged such that their inlet jets (of width *H*) are inwardly directed and symmetrically opposed. A top plate bounds both jets on one side of the inlets, while the other two walls curve away from each inlet jet before merging into a throat of width *W*. As in the standard virtual cyclone, the underlying principle of the OFVC is that each inlet jet follows its adjacent wall as it curves away, creating a recirculation region between the jets and the top plate. Because of centrifugal forces, particles move away from the wall and towards the plane of symmetry. These particles eventually cross from the main flow, becoming concentrated in the central recirculating zone. Eventually particles escape the recirculation zone to be highly focused into a narrow region about the plane of symmetry between the two converging flows. A small flow could also be provided through a porous top plate for the purpose of purging particles from the recirculating region, eliminating any time delay before particles escape the recirculation region.

Simulations show that the OFVC will only concentrate particles within a specific size range. Too small particles will tend to follow the main flow instead of moving centrifugally across the dividing streamline. Too large particles will overshoot the recirculating region and pass into the opposing jet, and thereby resist focusing. A closed-form result for the focused particle size was derived based on an approximate analysis of an OFVC. The key parameters controlling the focused size range are the inlet width and velocity, total turning angle, radius of curvature (of the curved walls), throat width, particle density, and gas viscosity. The analysis showed that smaller particles can be concentrated by decreasing the inlet slit width, or by increasing the inlet gas velocity or the turning angle. This has interesting implications for micromachined devices, as the fabrication of OFVC devices with very thin slits would enable concentration of very small-sized particles (for a fixed inlet velocity), or of larger particles with very low inlet velocities. The ability to accomplish particle concentration at low gas velocities would greatly reduce the pressure drop (and power consumption) through such devices.

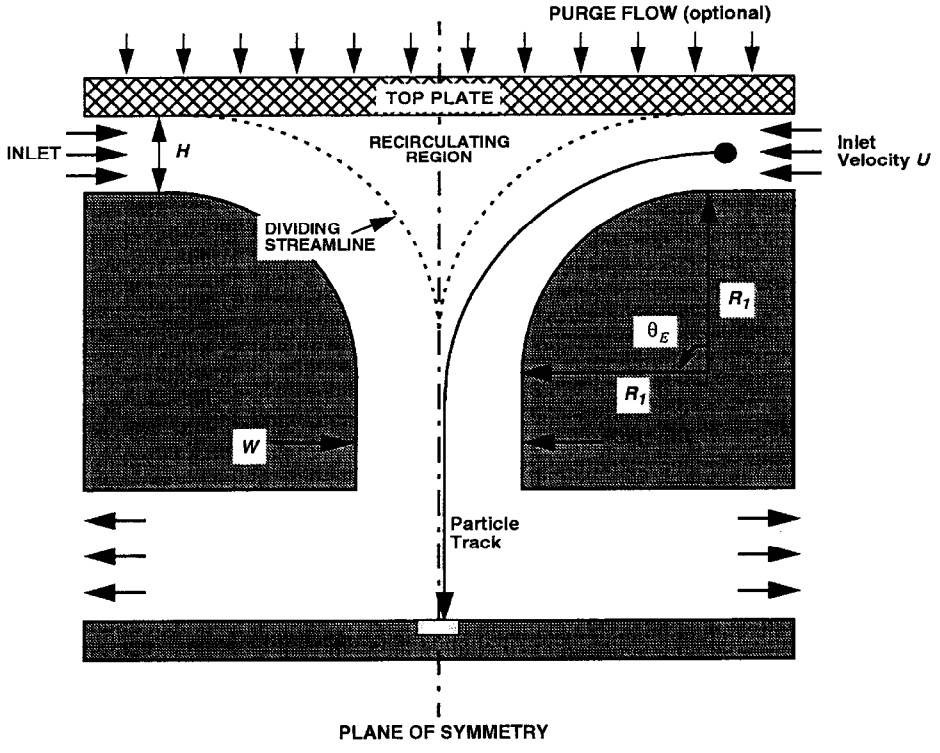

Figure 3. Schematic diagram of the opposed-flow virtual cyclone.

As an example, consider the design of an OFVC operating with air at atmospheric conditions. To avoid overly small physical dimensions, the device should be operated at as high a velocity as possible (but not so high as to raise compressibility concerns). For this example, consider the 2-D OFVC shown in Figure 3 with $W=2H$, and $\theta_E=\pi/2$, and assume that particles of interest are in the size range from 1.5 to 5.0 micron, i.e.

respirable particles. Numerical examples using the closed-form analytic result are given in Table 1 for some typical velocities/pressures. The flow rate per unit length of collector, Q/L, is found from the relation $Q=2UHL$ where the factor of two reflects the two inlet slits per OFVC. As seen in the table, operation at low pressure drops requires very small slit widths. Typically, the highest velocity up to which compressibility effects can be ignored is U=100 m/s; at this velocity respirable particles can be concentrated with a slit width slightly larger than a millimeter.

Table 1: Examples of possible OFVC designs to concentrate/collect particles in the 1.5 to 5 μm range for various gas inlet velocities (based on the 2-D OFVC geometry shown in Figure 3).

U (m/s)	$\sim\Delta P$ (torr)	H (mm)	W (mm)	R_1 (mm)	Q/L (lpm/cm)
10	0.45	0.12	0.24	0.67	1.44
20	1.81	0.24	0.48	1.34	5.76
50	11.3	0.60	1.2	3.36	36.0
100	45.1	1.2	2.4	6.71	144.

Numerical finite-element flow simulations (FIDAP 7.60) were performed to test the validity of the OFVC concept and the above design equations. The design goal for this example was to collect/concentrate a bioaerosol (of particle density 1 g/cm^3) in the size range from 1.5 to 5.0 μm with a low pressure drop. The following conditions from Table 1 were assumed for these calculations: U=50 m/s, H=0.6 mm, W=1.2 mm, and R_1=3.5 mm. A small purge flow with a velocity of 0.5 m/s was introduced through an assumed porous top plate to avoid development of a large recirculation flow. Some simulation results are shown in Figure 4. Note that because of symmetry, each of the four plots is used to show two results, one on the left-hand side and one on the right-hand side.

As shown in Figure 4a (left side), the calculated fluid streamlines stay attached to the lower wall as it curves. Particle trajectories are shown in Figure 4b, 4c, and 4d; note that particle trajectories which cross the plane of symmetry are specularly reflected back so that the entire path can be shown on one side of the plot. Particle behavior was found to be in good agreement with the analytic predictions: 1.5 μm particles showed moderate focusing, particles between 2 and 5 μm showed excellent focusing, and focusing was lost for particles larger than about 5 μm. The extent of particle focusing along the plane of symmetry is very high for particles in the range of 1.5 - 5 μm.

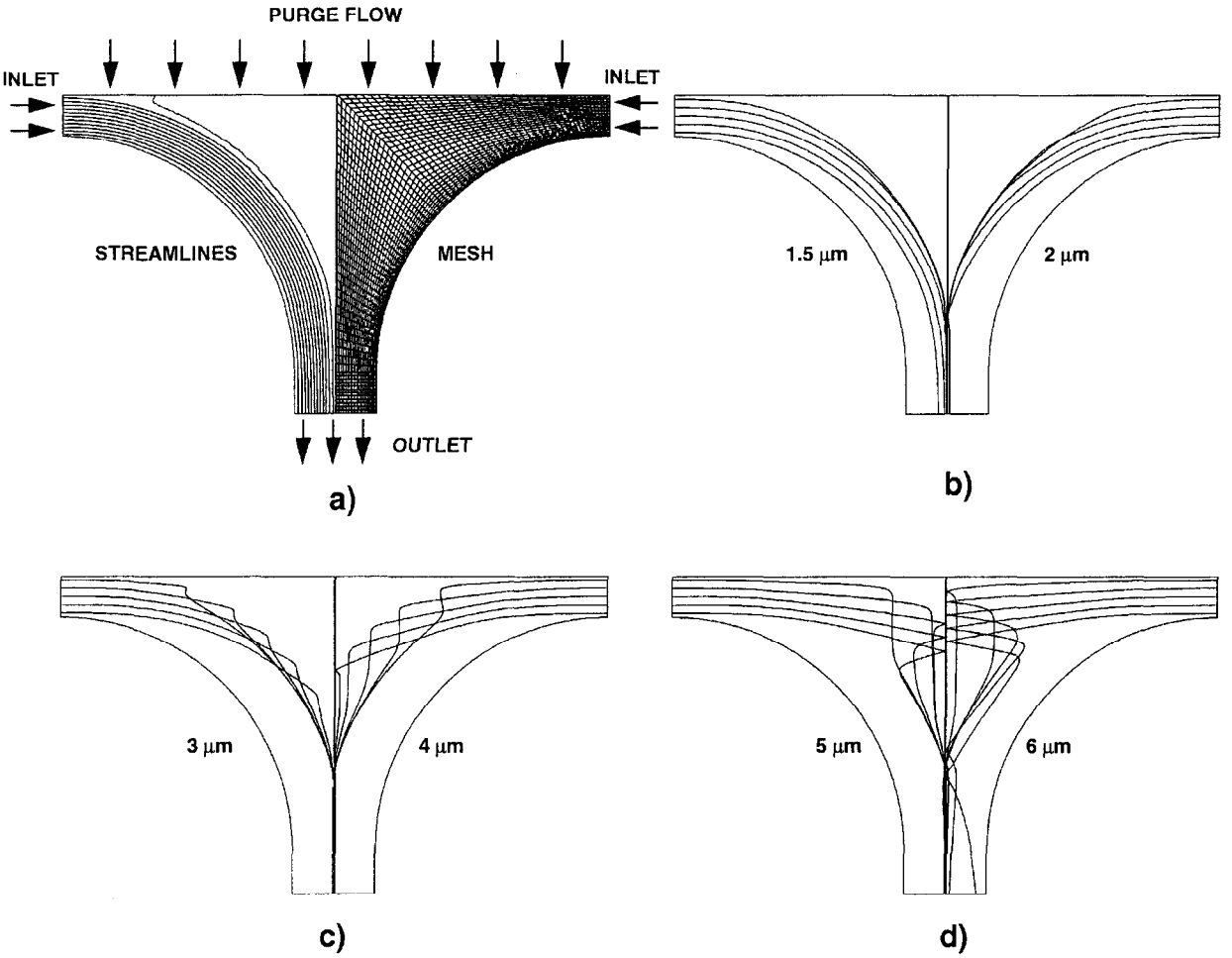

Figure 4. Flow Simulation Results Using FIDAP 7.6

Results for flow streamlines and particle trajectories are shown based on numerical fluid/particle simulations of a 2-D opposed-flow virtual cyclone:
a) flow streamlines (left) and mesh (right); particle trajectories for: b) 1.5 μm (left) and 2 μm (right), c) 3 μm (left) and 4 μm (right), d) 5 μm (left) and 6 μm (right).

As both the virtual cyclone and the OFVC concepts were originally developed based solely on numerical simulations, it is crucial that the model predictions be experimentally verified. A previous experimental program has verified that the qualitative behavior of the flow and particles in the virtual cyclone matches model predictions [2]. Experimental verification of the OFVC has been conducted as part of this LDRD project. An OFVC test device similar to that shown in Figure 3 has been constructed based on a 400 micron throat width. The test device is modular so that a variety of inlet widths can be tested. The characterization experiments were performed with monodisperse particles generated by a commercial device (TSI Vibrating Orifice Aerosol Generator). Particle transport through the OFVC was measured with an Aerodynamic Particle Sizer, which allowed independent verification of particle size and concentration. Initial results confirm the qualitative behavior of the OFVC. We hope to conduct more comprehensive performance tests on

prototype OFVC micro-separators during FY2001.

Based on the analysis and numerical simulations performed under this LDRD, a patent application for the OFVC has been filed with the United States patent office (SD-8192).

Focusing 2-Dimensional Wedge Impactor

An alternative micro-separator design concept is shown schematically in figure 5. This concept, referred to as the *focusing 2-dimensional wedge impactor*, performs a slightly different function from that performed by the *opposed-flow virtual cyclone*. The OFVC is designed to collect particles within a range of particle aerodynamic diameter sizes which can be selected by the designer. The focusing 2-D wedge impactor is designed to capture all particles larger than a minimum aerodynamic diameter. Particles smaller than this diameter stay suspended in the main flow and thereby simply pass through the system. Particles larger than this minimum diameter are first focused into a concentrated stream by the focusing wedge geometry, and then captured via inertia forces which carry the particles into the impinger aperture while the main flow is suddenly diverted through a 90 degree turn into the outlet flow channels on either side of the impinger aperture, parallel to the surface of the collection plate. This performance may be better imagined by referring to Figures 5, 6 and 7 below.

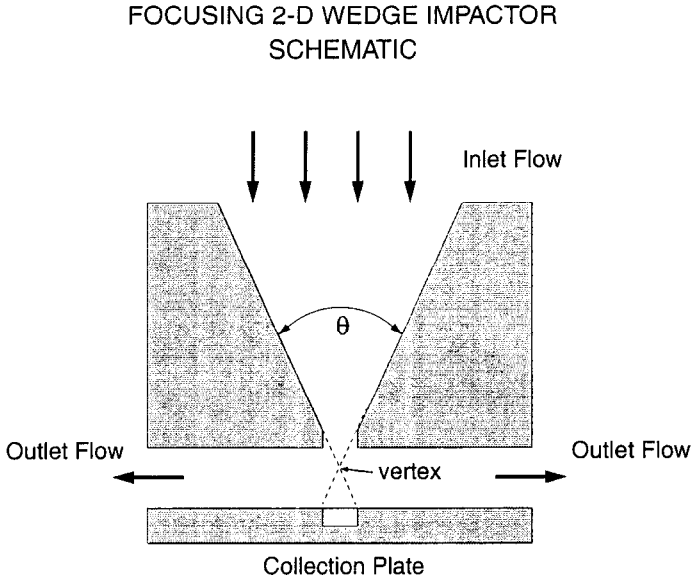

Figure 5. Schematic of the Focusing 2-Dimensional Wedge Impactor Concept

As mentioned, the Focusing 2-D Wedge geometry shown in Figure 5 allows for the concentration of particles larger than a minimum size. This capability became important because some potential users of our aerosol concentrator insist on the ability to sample

the entire particle burden rather than only particles in the respirable range. The calculated streamlines of Figure 6 show how this geometry would concentrate particles into an impinger aperture centered about the plane of symmetry. Also, this geometry can be simpler to fabricate than the OFVC because there are no curved surfaces nor does one need to deal with the issue of a recirculation zone. On the other hand, the geometry is incapable of excluding larger particles as can the OFVC. Therefore, any mating micro-impinger must be able to deal with the greater particle burden and larger particles.

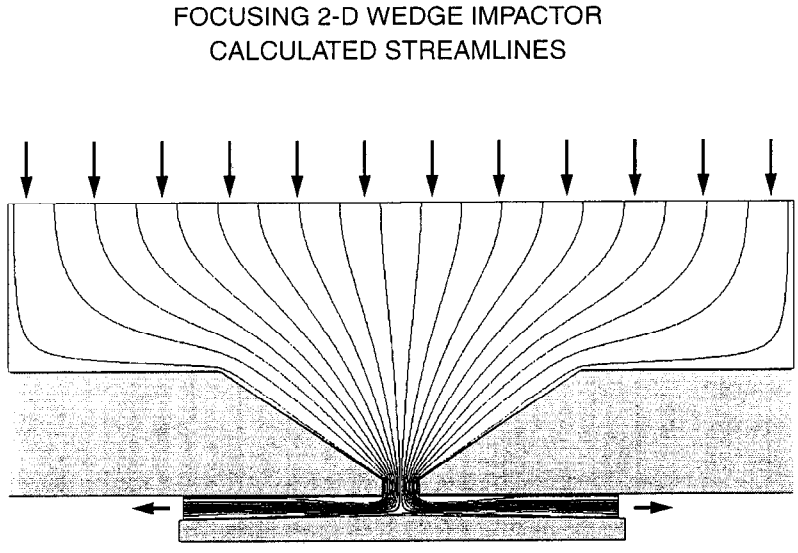

Figure 6. Calculated Streamlines Illustrating the Operation of the Focusing 2-D Wedge Impactor

Figure 7 shows how the Focusing 2-D Wedge geometry deals with particles in the respirable size range. As can be seen from the figure, the geometry is optimally focused for a specific particle size. Smaller particles are not quite as well focused, while larger particles will exhibit a broader range of dispersion. A well chosen impinger aperture can be selected to essentially insure that all respirable particles are collected. However, as mentioned above, there is no way to exclude the impingement of a certain percentage of the larger particles in the incident particle burden, since those that enter the collector near parallel to and very near the plain of symmetry are likely to impinge on the impinger aperture. It would require further experiments to determine under what circumstances of overall particle burden or particle size this would become a problem for any specific micro-impinger design.

Given an appropriate impinger, the Focusing 2-D Wedge Impactor geometry can be remarkably effective. Figure 8 presents experimentally determined collection efficiencies for a wedge geometry designed to collect particles of 1.5 micron particle diameter and larger. The wedge has an included angle of 110 degrees and a throat width of 430

microns. Testing was done using essentially the same configuration as previously describe for testing the OFVC. As can be seen from the figure, a sharp transition can be achieved from near-zero collection efficiency below the cut-off particle size to a very high collection efficiency, 85% or greater, above the cut-off particle size. The exact cut-off particle size is effected by the velocity of the air flow through the separator, but can be remarkably sharp nonetheless. In the experiments of Figure 8, a velocity increase of one-third reduces the cut-off particle size by only about 0.2 microns. The transition is easily complete within a range of particle diameters of one micron or less, with the bulk of the transition easily complete within half or less of that range.

Figure 9 shows results of experimental pressure drop through the Focusing 2-D Wedge Impactor compared with the approximate theoretical pressure drop estimated from the energy of the flow. A FIDAP simulation calculation provides a third check at a single point. The results show very good agreement between all three methods. This suggests that the approximate theory is an acceptable method for conveniently estimating the pressure drop performance of this geometry.

Micro-Separator Fabrication

The results from Table 1 would suggest that the smaller one can make the features of the micro-separator, the lower the velocity required and the lower the pressure drop and power consumption that can be achieved. This could have powerful implications for a micro-separator, especially for configurations not constrained by the needs of a liquid micro-impinger. In a system with a liquid free-surface micro-impinger, as we have here, certain limitations of the liquid micro-impinger come into play long before micro-fabrication limitations are reached for the micro-separator. Simply stated, the micro-flow channels of a micro-impinger cannot be made ever smaller because the liquid channel depth and volume need to be sufficiently large to avoid serious perturbation due to impingement of the respirable particles. While it is possible to model impingement in detail, for practical application, a conservative performance estimate may be had by keeping the particle diameter at 5% or less of the width and depth of the channel. Thus, for respirable particles, a micro-flow channel on the order of 100 microns deep by 100 microns wide might represent a prudent lower limit. This channel width is a good match to an OFVC micro-separator geometry with a throat width of about 400 microns. This value was used along with the design equations derived earlier to generate the OFVC cross-section used for the micro-separator shown in Figure 10. This micro-separator consists of an array of six parallel OFVCs each approximately 20 mm long. The complete array fits within a 20 mm by 20 mm square. The micro-separator has been conventionally machined from PEEK (polyetheretherketone) engineering plastic using a miniature contour tool which, in its turn, is produced using conventional miniature machining techniques. Smaller OFVC geometries would require contour tools generated using more extraordinary miniature machining techniques, for example, micro-WEDG (wire electric discharge grinding), or ion beam machining for even smaller tools. Micro-molding or micro-embossing are also suitable fabrication techniques, once the master tool or mold has been made.

FOCUSING 2-D WEDGE IMPACTOR CALCULATED PARTICLE TRACKS

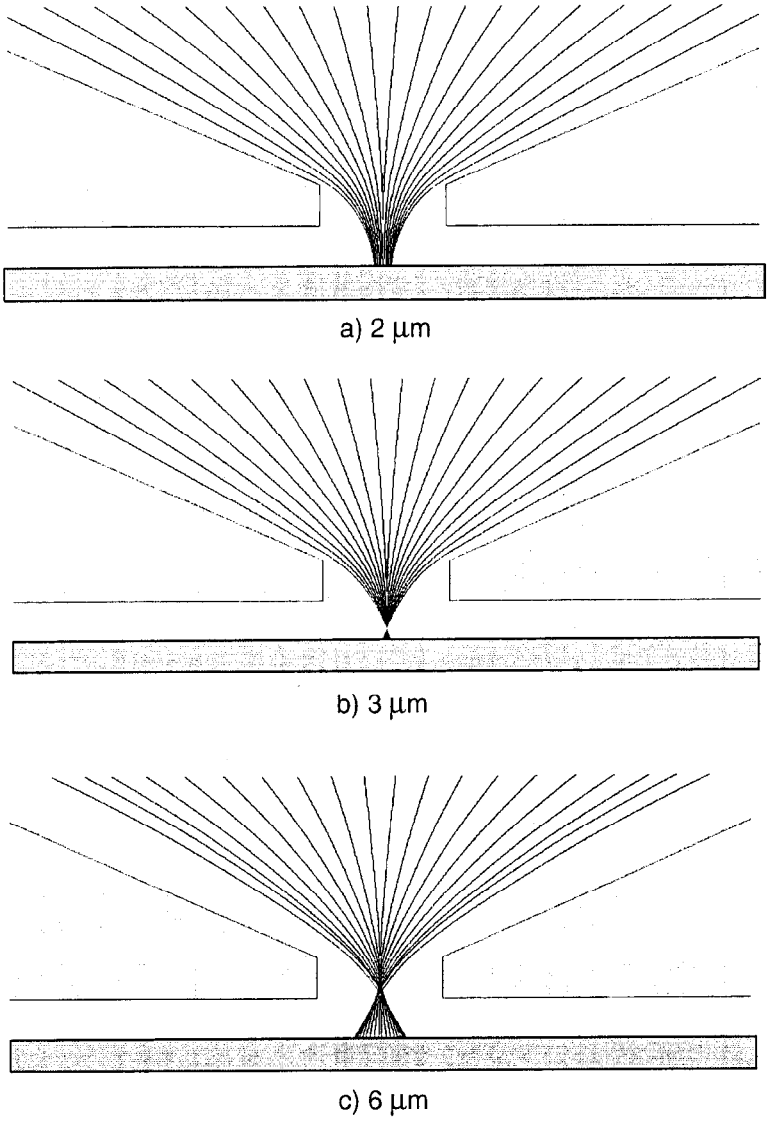

Figure 7. **Focusing 2-D Wedge Impactor Calculated Particle Tracks. Figure 7a) shows the trajectory for particles of 2 micron diameter; Figure 7b) shows the trajectory for particles of 3 micron diameter; Figure 7c) shows the trajectory for particles of 6 micron diameter.**

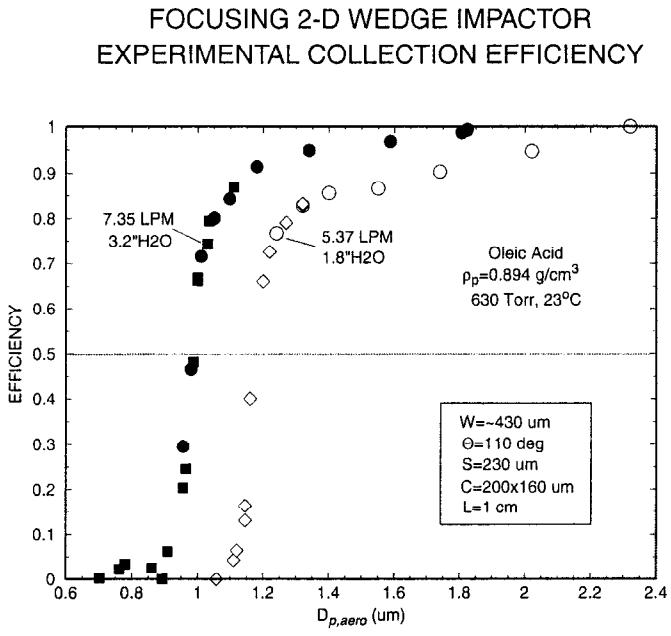

Figure 8. **Collection Efficiency of a Focusing 2-D Wedge Impactor nominally designed to collect particles of 1.5 micron particle diameter and larger.**

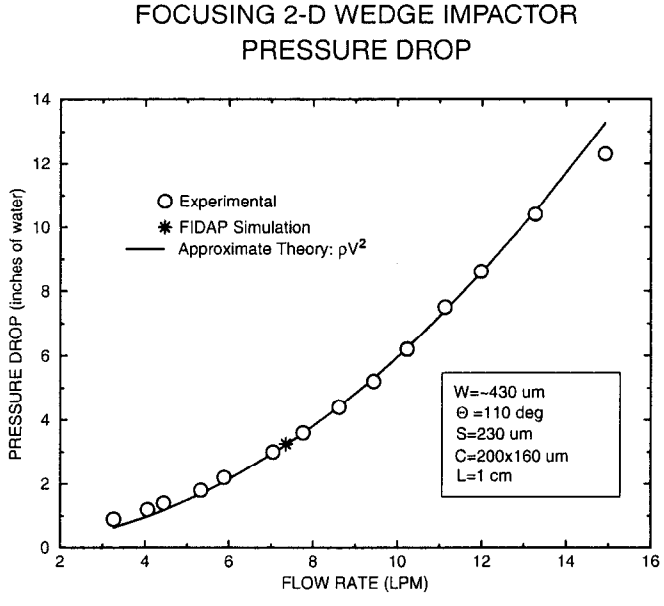

Figure 9. **Flow Resistance for the Focusing 2-D Wedge Geometry. Comparison between experimental performance, FIDAP simulation, and Theory.**

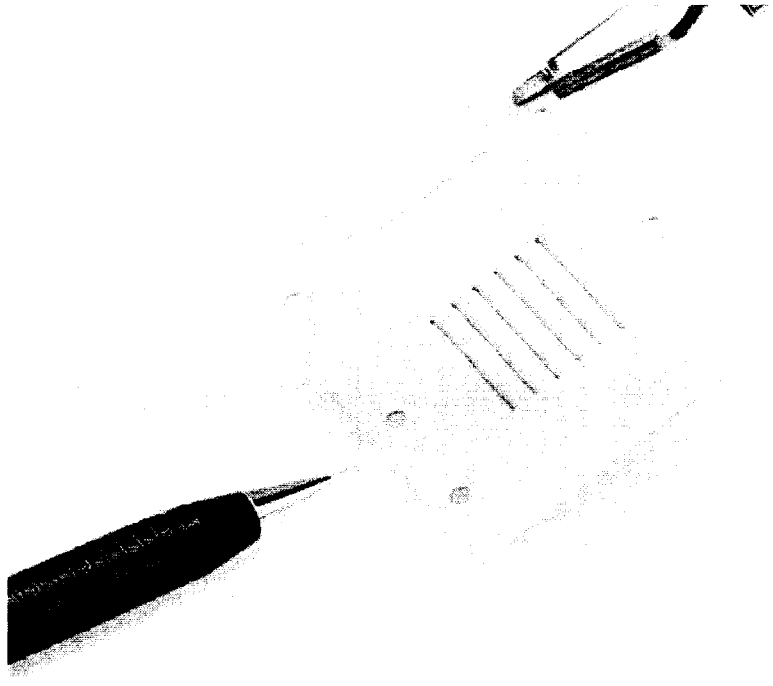

Figure 10. Photograph of a Micro-Separator Plate containing an Array of Six Slots with Opposed Flow Virtual Cyclone Cross-Section.

Micro-Impinger Development

In many ways, the micro-impinger portion of the miniature bioaerosol concentrator is the most challenging element in the entire concept. For compatibility with aqueous-based microanalysis and detection schemes, particles must be concentrated into a liquid sample volume of about one microliter (or less). For a miniature battery-operated device, gas sampling is likely to remain on the order of 10 – 20 liters per minute, which is the approximate air moving capacity of the smaller DC-powered mini-blowers, developed for component cooling in electronic applications, that we hoped to use for our concentrator. This is a very small rate for sampling BWA bioaerosols. A sample of 10 – 20 liters per minute taken from a BWA bioaerosol may collect only ten to twenty active particles over the course of one minute. Unless longer sampling periods are employed, this yields so few active particles to be separated, concentrated, analyzed, and detected, that the performance of both the bioaerosol concentrator and the microanalysis system will need to be first rate for reliable detection.

More conventional approaches to the particle separator/impinger function would be a poor fit with microanalysis systems, which use microliter (or less) sample volumes.

These approaches, which include liquid droplet cyclones, inertial separators with dropping film impingers, electrostatic precipitators that are periodically washed down, etc., all tend to require grossly too much liquid when viewed from the perspective of a one microliter sample volume. And while it might be possible to perform post-collection particle concentration from milliliter liquid sample volumes down into microliter liquid sample volumes, we could see no reliable techniques that would produce the required thousandfold particle concentration and liquid volume reduction and still be achievable in a low-powered miniature package. Therefore, we elected to attempt the approach of direct impingement into microliter sample volumes. This approach puts the burden of particle separation and focusing onto a micro-separator. The micro-impinger then needs to present a suitable free surface to the micro-separator for impingement, and yet still incorporate a means to move the fluid sample through the micro-impinger, all-the-while constrained within a sample volume limited to one microliter. We felt this could best be accomplished by micro-flow in open channels, with liquid motion in the open channels derived from electro-osmotic flow. The concept was shown schematically in Figure 1.

Electro-osmotic flow is a convenient way to achieve uniform liquid motion in open micro-channels. In fact, few other options present themselves. Pressure flow is impractical with open channels. Gravity flow would impose a major operational and handling constraint upon the system. Suction flow might be possible, although it, too, seems fraught with difficulties in an open channel system. Electro-osmotic flow imposes certain constraints upon the materials used in the system. Ideally, the channels of the micro-impinger would be of an *ionogenic* material such as glass (bare metals are precluded). Other dielectric materials have also been shown to work, including ceramics and plastics. From time to time, attempts have been made to use metal structures coated with a dielectric. This approach recommends itself because it would be compatible with silicon microfabrication techniques. Sandia, too, investigated this approach as part of its microChemLab development activities. To date, all such efforts to use microfabricated metals (i.e., *silicon*) as an underlying structure have proved unsuccessful because the dielectric coatings that have been applied have not been able to withstand the combination of aqueous environment and high voltages which the electro-osmotic flow entails. Dielectric breakdown generally ensues in short order, often within days if not hours. (Parenthetically, the micro-separator, too, needs to be of dielectric material to avoid electrical arcing from the high voltage potential on the aqueous electrolyte in the micro-channels. The micro-separator is within fractions of a millimeter separation distance from the electrolyte. This would easily arc-over if the micro-separator were metal. We make the micro-separator of PEEK engineering plastic.)

Electro-Osmotic Flow

If a potential gradient is applied along the length of liquid in a dielectric micro-channel, the liquid will flow due to electro-osmosis [4]. The velocity will depend on the potential between the liquid adjacent to the channel wall and the liquid far away from the wall. This potential, the zeta potential, ζ, is a function of the liquid and the wall material. In

symbolic form, the velocity, v, will be proportional directly to the applied potential, E, and zeta potential, ζ, and inversely to the liquid viscosity, η:

$$v \propto E\zeta/\eta$$

While this relationship is usually derived in the literature for flow in a cylindrical channel, the proportionality relationships hold for electro-osmotic flow in open channels. Here, the motivating force is exerted at the channel walls (bottom and sides for a rectangular channel), and the free surface provides essentially no resistance to flow along the direction of the channel. The established flow will approximate plug flow, so characteristic of electro-osmotic flow.

Serpentine Micro-Impinger Configuration

The micro-impinger configuration that comes most directly to mind is one long micro-flow channel centered under one long OFVC micro-separator. The geometry of the micro-impinger channel and the OFVC micro-separator are interrelated by the double constraints imposed by the selected air flow and the selected microfluidic collection volume. A micro-separator suitable for sampling 10-20 liters of air per minute can be obtained by choosing a throat width of approximately 400 microns and an overall length of approximately 110 millimeters. An idealized micro-channel of corresponding length might have a channel width of 75 microns (centered on the 400 micron throat), and a depth of 125 microns, and will contain a liquid volume of about 1 microliter, which is our target volume. One way to arrange this in-line configuration so as to center it over a commercially available mini-blower is to arrange it into a serpentine traversing back and forth across the face of the blower. This is the origin of the geometry used in our design.

A micro-impinger exhibiting this geometry can be seen in Figure 11. The micro-impinger is a two layer diffusion-bonded sandwich made of fused silica. The sandwich construction allows us to have closed micro-flow channels in areas not directly involved with particle impingement. Fused silica, although more difficult to diffusion bond than the borosilicate glass with which we prefer to work, is used to enable the laser micromachining of the thru-slots on either side of each micro-channel. The borosilicate glass does not lend itself to laser micromachining. These through-slots allow the mini-blower to draw air through the micro-impinger and micro-separator in the proper flow pattern. In this example, the straight traverse legs of the serpentine are open channel sections, while the feed and collection channels and the reverse bends are all closed channel sections.

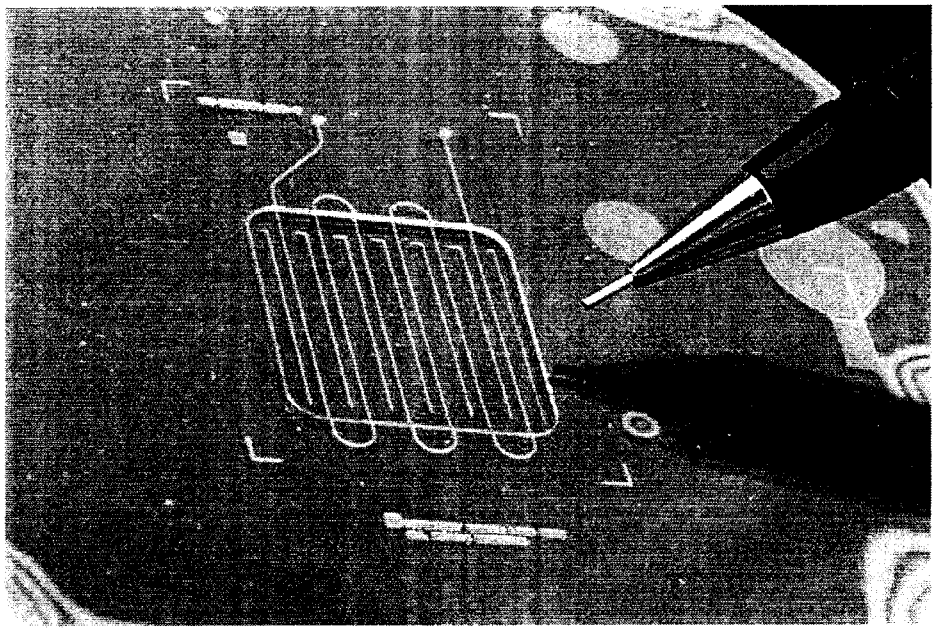

Figure 11. Photograph of a Micro-Impinger Wafer Sandwich prior to dicing. One can see: a) the large aperture window exposing the parallel open channel segments of the serpentine micro-flow channel; and, b) the laser-machined thru-slots to either side of each traverse of the micro-flow channel.

The serpentine micro-impinger has one inherent drawback. The serpentine's long length requires a higher applied voltage potential from its inlet to its outlet to drive the fluid at a given velocity. Using aqueous electrolytes as used in microanalysis systems, flow velocities of about 1 millimeter per second are obtained for applied potential gradients of 20 volts per millimeter. To achieve this performance in the serpentine, the total applied voltage would exceed two thousand volts.

One inherent characteristic of the serpentine that some may consider a desirable feature is that each subsequent traverse of the serpentine collects additional particles, so that the highest concentration of collected particles is achieved at its outlet.

Problems with the as-fabricated Micro-Impinger

We encountered a number of unexpected problems with the as-fabricated micro-impinger which have greatly impeded our development efforts. They have, so far, prevented us from achieving acceptable micro-flow performance in our micro-impingers.

Our idealized micro-impinger was conceptualized around photo-etchable glass. In our idealized micro-impinger, we would have preferred a narrow, deep micro-channel. Channels of this tall aspect ratio are not fabricable in conventional glass via photo-etching. Initially, we had hoped to achieve the desired tall aspect ratio by making our

micro-impingers from Foturan photo-etchable glass/ceramic. It is possible to etch deep channels as narrow as 50 microns in Foturan. It would also be possible to produce the thru-slots by etching. However, early examples of micro-channels in photo-etchable glass produced structures which were reported to have inconsistent electro-osmotic flow performance from example to example. There were also questions related to diffusion bonding. Our materials advisers urged us to abandon photo-etchable glass.

Our only readily available alternative was to fabricate the micro-impingers from fused silica glass. Fused silica could be etched, could be diffusion bonded, and would enable laser micromachining of through-slots. Unfortunately, no glass material can be photo-etched to produce deep, narrow micro-channels. The micro-channels that were produced are characteristic of etched glass, approximately twice as wide, at 200 microns, as they are deep, at approximately 100 microns. This fourfold increase in micro-channel width over our original idealized concept would introduce problems that we did not adequately appreciate at the time of this substitution of material and fabrication technique.

These problems became apparent soon after we began working with these micro-impingers. Filling of the micro-channels was often problematic because of wicking performance. The as-fabricated wide aspect ratio micro-channels have only about one-fourth the wicking capability of the idealized design. This weaker wicking capability, coupled with a subtle and not fully appreciated wicking break where closed channels transitioned to open channels, resulted in micro-impingers that were not self-priming by capillary action, and had to have separate channel sections filled by hand.

Evaporation from the open channel sections of the micro-impinger is the major problem with the as-fabricated micro-impingers. The as-fabricated micro-channels have four times the free surface area of the idealized design, and, therefore, at least four times the evaporative loss. This problem is aggravated by the fact that the as-fabricated micro-channels have only about one-fourth the wicking capacity of the idealized design. Coupled with the unintended wick breaks at points of transition from closed channel to open, wicking performance was sufficiently erratic so as to be unable to keep up with the higher rate of evaporation. For the typical aqueous electrolytes we would normally use, evaporative losses and priming problems were too severe for the micro-impingers to function as required.

We are considering a number of possible correctives. To address the evaporation issue, one possible corrective is to use lower volatility electrolytes. This potentially would be the readiest solution to evaporation, but has its own faults. Unfortunately, most liquids that exhibit lower volatility than water also have a significantly higher viscosity. The electro-osmotic flow velocity is inversely proportional to the liquid viscosity, so these liquids may have significantly slower flow rates than we would prefer. A proportionately greater voltage potential could be applied to restore the desired velocity, but this may not be a solution one would prefer. A second corrective might be to revert to the idealized design by exploring alternative fabrication materials and techniques. Plastic micro-impingers fabricated using micro-molding or micro-embossing may be feasible.

Alternatively, we may revisit photo-etchable glasses to consider whether this material may have been abandoned too quickly. We hope to explore these alternatives in FY2001.

Another corrective may be available if we abandon the serpentine micro-impinger for a potentially less problem prone configuration.

Alternative Micro-Impinger Configuration

A configuration which might be less prone to the problems of the single serpentine channel is the parallel flow network shown in Figure 12. In this network, flow enters the network from a feeder channel at one corner of the network and leaves through a connector channel at the diagonally opposite corner of the network. Flow channels of identical cross-section make up the parallel array. In pressure flow systems, parallel flow arrays achieve uniform flow by sharing a common inlet manifold and common outlet manifold, each at its own common operating pressure. However, in an electro-osmotic flow system, uniform flow in the parallel channels must be achieved in a more complex manner. The appropriate constraint is that the parallel channels share a common uniform voltage potential gradient along their length, although actual voltage potentials only match at the points of flow divergence or confluence. This constraint requires that the flow cross-sections of the feeder (or collector) manifold must decrease (or increase) at each flow divergence (or confluence). The parallel array shown in Figure 12 exhibits this geometry. Calculated electro-osmotic flows, depicted by grey-scale shading in the figure, demonstrate that this arrangement can achieve uniform constant flow in all of the parallel channels of the network.

The calculated results represented by the grey-scale shading in Figure 13 illustrate how the voltage potential gradient naturally arranges itself as a result of this geometry. The voltage potential gradient along the length of each of the parallel channels is uniform and constant from channel to channel, while actual voltage potentials decrease from the inlet to the outlet side of the array, as dictated by the flow geometry. Voltage potentials at diverging or converging flows must match.

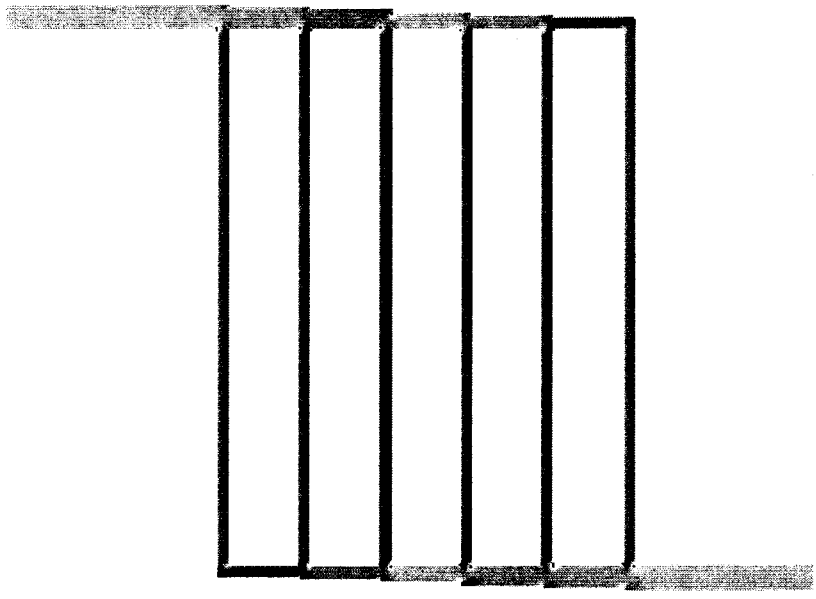

Figure 12. Calculated Electro-Osmotic Flow for the Proposed Parallel Flow Micro-Impinger. Flow enters the network from the lower right-hand channel and leaves by the upper left-hand channel. Grey-scale shading represents flow.

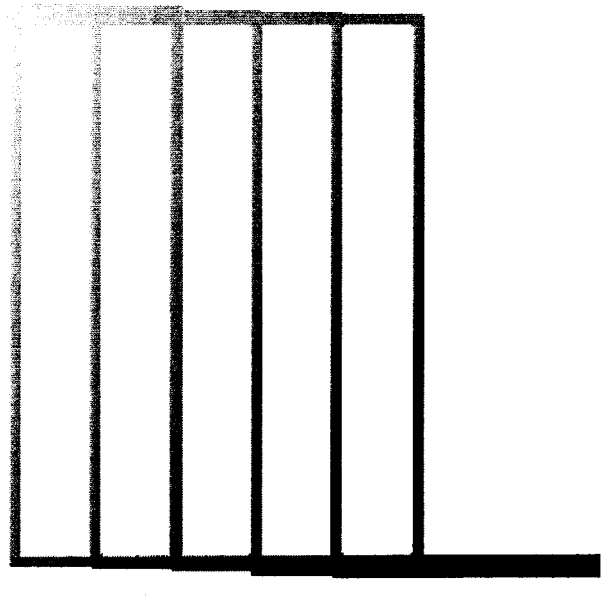

Figure 13. Calculated Electro-Osmotic Potential for the Proposed Parallel Flow Micro-Impinger. Flow enters the network from the lower right-hand channel and leaves by the upper left-hand channel. Grey-scale shading represents voltage potential.

These calculated results are encouraging. To examine this alternative further, micro-impingers with this parallel flow design were microfabricated in fused silica using photo-etching. Parallel flow networks require thorough development and testing because parallel flows are susceptible to flow upsets that do not occur in a single long channel. Although the calculated performance is good, real performance might be adversely influenced by unintended variations and imperfections as will occur in all real fabricated parts. It is important, therefore, to demonstrate that perturbations due to these variations will not effect the stability of the parallel flow network.

Miniature Bioaerosol Concentrator Assembly

The overall configuration of the miniature bioaerosol concentrator is shown in Figures 14 and 15. Collection liquid flows to and from the concentrator assembly through micro-capillaries terminated in miniature fittings. Liquid originates in a source reservoir (not shown, which we would anticipate would be part of the integrated microanalysis and detector assembly. This reservoir contains one electrode of the electro-osmotic circuit. Outflow is directed to the inlet of the microanalysis device to be served by the concentrator. A second electrode is teed into this outflow line to complete the electro-osmotic circuit.

The various layers of the micro-separator (top plate, OFVC plate), micro-impinger, and assorted spacer and housing plies are stacked in layer-cake fashion. The plies must adequately seal against each other to insure that air will be drawn from the intake to the system.

A commercially available mini-blower is mounted with its inlet facing toward this assembly. The blower draws air through the micro-separator and micro-impinger, and exhausts it out the side of the blower (see Figure 15). Electronics to control the blower, and electrical connectors, may be mounted to the printed circuit board/spacer ply, which is included for this purpose as part of the overall assembly stack.

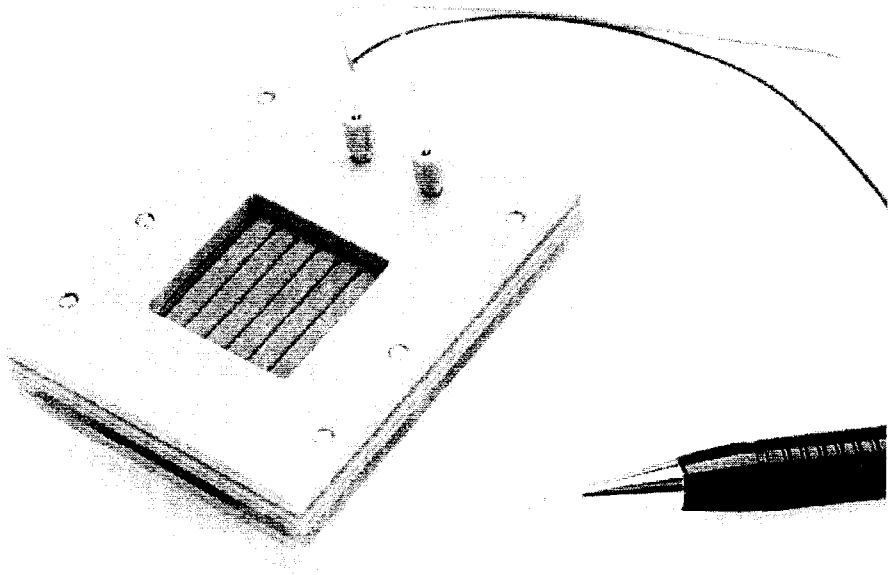

Figure 14. Miniature Bioaerosol Concentrator showing slots for air intake.

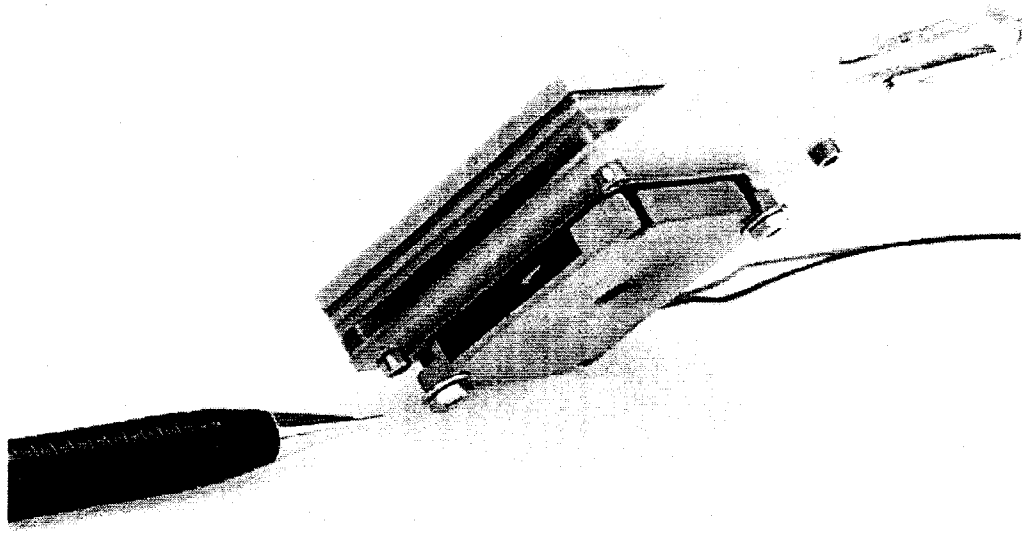

Figure 15. Miniature Bioaerosol Concentrator showing the mini-blower and the multi-plate assembly.

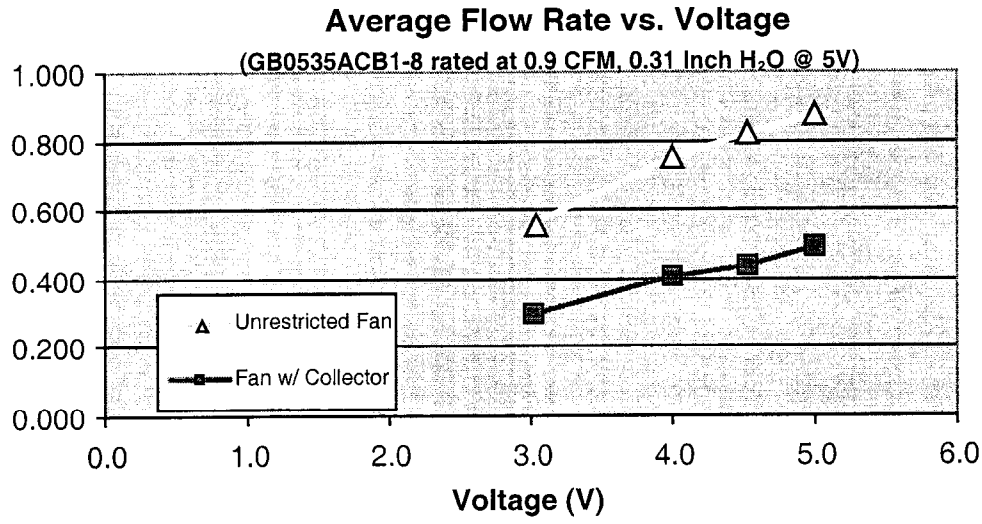

Figure 16. Mini-Blower performance measurements showing that a flow rate of 10 to 20 liters per minute ought to be achievable.

An important question is whether commercially available mini-blowers have the requisite performance needed to draw the desired flow through the miniature bioaerosol concentrator assembly. Figure 16 shows performance results obtained for a highly efficient mini-blower drawing air through a micro-separator assembly (top plate, OFVC plate) mounted into the miniature bioaerosol concentrator housing. No micro-impinger was yet available at the time of this testing. The micro-separator contains the most restrictive sections of the air flow configuration, the opposed flow inlet jets (of width *H* in Figure 3) and throat section (of width W=2H from Figure 3). The results indicate a flow of about 0.5 CFM (approximately 14 liters per minute) at a nominal 5 volts applied to the DC blower motor. The addition of the micro-impinger will further reduce this flow rate considerably, but performance is certainly within the correct range. If necessary, the DC blower motor can be run at a somewhat higher voltage than the nominal 5 volts specification. These motors are of a special, low power, high torque design with heavier gauge motor windings than are normally found in such mini-blowers. Overall power draw for the configuration of the experiment was approximately 0.4 watts.

The separation and focusing performance of the OFVC micro-separator is governed by the speed of the airflow through the inlet jets and throat, which in turn, is adjustable by control of the mini-blower speed via applied motor voltage. We assume that a final design configuration using standard interchangeable parts will yield a uniform performance from assembly to assembly, allowing a fixed performance to be established once the motor voltage corresponding to the desired jet velocity (e.g. to separate and focus respirable particles of 1 to 5 microns diameter) has been determined from prototype testing. Of course, uniformity of performance from device to device would need to be demonstrated for prototypes before this assumption could be confirmed. We do not anticipate active or adaptive motor control, although these control schemes are not

precluded by the design. Rather we anticipate a fixed voltage / fixed flow rate / fixed performance for the design.

References

1. Torczynski and Rader, 1997, "The Virtual Cyclone: A Device for Nonimpact Particle Separation," *Aerosol Sci. Technol.*, **26**: 560-573

2. Torczynski, O'Hern, Rader, Brockmann, and Grasser, 1998, "An Experimental Investigation of the Flow in a Virtual Cyclone," SAND98-2004

3. Patent Application SD-8192

4. Davies, J. T., and Rideal, E. K., "Interfacial Phenomena", 1963, Academic Press, New York and London.

Distribution:

Unlimited Release:

3	MS 9018	Central Technical Files, 8945-1
1	MS 0899	Technical Library, 9616
1	MS 9021	Classification Office, 8511/Technical Library, MS 0899, 9616
1	MS 9021	Classification Office, 8511 For DOE/OSTI
1	MS 0188	D. Chavez, LDRD Office, 1030
1	MS9004	J. Vitko, 8100
1	MS9951	D. Lindner, 8101
1	MS9104	W. Bolton, 8120
6	MS9103	K. Wally, 8120
1	MS9105	J. A. Lamph, 8111
2	MS0827	D. J. Rader, 9112