Development of an Orientation-independent Handheld Nebulizer

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Abstract

This work aims to develop and characterize a miniaturized handheld vibrating mesh nebulizer (MHVMN) that can be operated equally well in any orientation. The MHVMN developed and tested in the present study comprises a power supply-frequency generator unit, a vibrating mesh plate, and a liquid reservoir equipped with a flat end orifice tube (FEOT) or a traditional wick for liquid delivery. Since the study focused on optimizing the dimensions of the FEOT which is a unique application of capillary force for solution delivery, the operating parameters including gap between reservoir wall and FEOT (L_LR-FEOT), diameter of orifice on FEOT (D_orifice), nebulizer orientation, and aperture size were examined to investigate their effects on the performance of the MHVMN. The output rate and residual of a nebulizer were determined gravimetrically. An aerosol size spectrometer (Welas 3000) and a Micro orifice uniform deposit impactor (MOUDI) were employed to measure both aerosol concentration and mass distribution of the generated particles. The developed nebulizer was mainly evaluated with deionized water and 0.9% sodium chloride solution. The results showed that the L_LR-FEOT of 0.225 mm was the optimal design because of the output rate and residual were similar for different orientations. The D_orifice needed to be larger than 0.8 mm, to guarantee the continuity of nebulization, and residual less than 4%. The FEOT-based MHVMN had significant advantage over the wick-based one, for much lower residual, which was important when the drug or solution was precious. The most significant and unique feature of the optimized MHVMN was the performance almost orientation independent. This made it ideal for aerosol medications, especially for patients lying on bed.

Keywords: Vibrating mesh nebulizer; Orientation-independent; Flat end orifice tube

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INTRODUCTION

For several decades, nebulizers, convert liquid drugs into tiny airborne particles, have been used for medication through inhalation (McCallion et al., 1996; Dhand, 2002; Martin and Finlay, 2015; Mansour, 2018). Aerosol therapy through nebulizers is a well-established method for treatment of patients with pulmonary diseases, including, but not limited to, asthma (Zimo et al., 1989; O'Driscoll et al., 1992; Demoly et al., 1998; Hess, 2008), chronic obstructive pulmonary disease (COPD) (O'Driscoll et al., 1992; O'Driscoll, 1997; Alhaddad et al., 2015) and cystic fibrosis (Bos et al., 2016; Baravalle-Einaudi et al., 2017) in both hospital and home settings. The role of nebulizers in treating acute lung injury (Tuinman et al., 2012), ventilation associated pneumonia (Rouby et al., 2012), dyspnea (Boyden et al., 2015), as well as for systemic administration of macromolecules (Siekmeier and Scheuch, 2008) has also been reviewed. As a consequence of an aging population and increased occurrence of respiratory diseases (Maio et al., 2006; Burney et al., 2015; Khakban et al., 2017), the market for nebulizers is expected to grow (Mansour, 2018; Pritchard et al., 2018). There exist many types of nebulizer, including jet, ultrasonic, and vibrating mesh nebulizers, with many commercially available models (Martin and Finlay, 2015; Tashkin, 2016; Mansour, 2018). The jet nebulizer was developed according to the Bernoulli principle to provide suction by forcing drug solution through a constriction and to reduce the liquid into a fine spray. Early jet nebulizers were clumsy, noisy, and inefficient due to high amounts of drug left in nebulizer reservoir (referred to as residual) (Rau, 2002; Fischer et al.,
2009). Although new jet nebulizers either utilize baffles that remove large droplets or have breath-enhanced aerosol production that leads to less drug loss during expiration to improve drug delivery efficiencies, the generated particle size distributions can vary significantly due to the changes of temperature and concentration of the nebulizing solution over the nebulization time (Phipps et al., 1990). Ultrasonic nebulizers incorporate a piezoelectric crystal vibrating at high frequencies to generate aerosol particles (Ari, 2014). Early ultrasonic nebulizers have many limitations compared with jet nebulizers, such as large residual, degradation of heat-sensitive materials (Taylor and McCallion, 1997; Watts et al., 2008), and not suitable for atomizing suspensions and viscous solutions (Taylor and McCallion, 1997; Elhissi and Taylor, 2005).

Vibrating mesh technology is a significant innovation for drug delivery. A vibrating mesh nebulizer (VMN) is composed of a piezo mesh disk, a piezo mesh driver, a circuit board with batteries, and a liquid reservoir unit. The piezoelectric material vibrates at a very high frequency, pushing the liquid drug through the mesh to form aerosol particles (Dhand, 2002; Vecellio, 2006). Compared with other types of nebulizers, VMNs offer many advantages, such as ease of use, high portability, low power consumption, low noise during operation and high efficiency with minimal residual (Waldrep et al., 2007; Waldrep and Dhand, 2008; Pitance et al., 2010). The details of new types of VMN have recently been reviewed (Pritchard et al., 2018).
Most VMNs are designed to work with a liquid drug reservoir aside, normally placed in the vertical position. For some current commercially available models, the position of nebulizer is an important factor affecting run time and variability in particle distribution (Skaria and Smaldone, 2010). Although it has been shown that the dose delivery of some VMNs were not significantly affected by orientation angle (Hardaker and Hatley, 2010), they are not allowed to tilt over 45º off vertical. In fact, medical nebulizers currently available are sensitive to orientation. When a nebulizer is operated in improper orientation, drug aerosols may be prevented from being generated as the solution inside the nebulizer cannot be effectively transported; it might also lead to spillage of the solution. Therefore, it will be ideal if solution transport mechanisms, such as capillary force, can be employed to carry the liquid drug, thus reducing or eliminating the effect of nebulizer orientation.

Capillarity described below is the phenomenon of a liquid rising or falling inside a narrow tube without external force. It is the result of adhesion and cohesion between liquid molecules and the capillary material. The relative strength of adhesion and cohesion affect the contact angle, \( \theta \), which is the angle between the free surface of a liquid and the capillary edge. The liquid surface will be raised if the angle is less than 90º. The height of liquid rise, \( H \), has been mathematically derived (Batchelor, 1967), given as

\[
H = \frac{2\gamma \cos \theta}{\rho g a}
\]
where $\gamma$ is the surface tension, J m$^{-2}$; $\theta$ is the contact angle between the liquid surface and wall, degree; $\rho$ is the density of the liquid, kg m$^{-3}$; $g$ is the acceleration, m s$^{-2}$; and $a$ is the capillary radius, m.

To make the devices more user-friendly, especially for patients lying on bed, it will be ideal to remove the nuisance restriction of nebulizer orientation. Therefore, the main goal of this work is to develop and improve the solution storage and delivery unit and make the vibrating mesh nebulizer more compact, portable, quiet, and operable in any orientation.

**METHODS**

The MHVMN consists of two major parts: the power supply-frequency generator unit and the vibrating mesh-liquid delivery unit, as shown in Fig. 1. The MHVMN is powered by a rechargeable lithium-ion battery (DC 3.7 V). The power consumption of this device is about 0.93 watt. The external dimensions of the MHVMN are $6 \times 4 \times 4$ cm$^3$, with weight of 60 g. The stainless vibrating mesh plate is 40 $\mu$m thick and 16 mm in diameter, and is mounted on top of the solution reservoir by using a fixing cap (Fig. 1(A)). The technical specifications of the mesh plates provided by the manufacturer are presented in Table 1. There are 2375 apertures on each mesh plate and the distance between neighboring apertures ($D_{ap-ap}$) is 160 $\mu$m. The aperture is tapered. The narrower end of the two aperture sizes used in this work are 6.5 and 9.5 $\mu$m. The volume of the home-made solution reservoir is about 3.3 ml ($\phi 9.3 \times 35$ mm$^2$). A unique flat end
orifice tube (FEOT) or traditional polyester (PE) porous materials (Fig. 1(B)) can be inserted into the reservoir to deliver solution by the force of capillary action. In present study, FEOTs with different outer diameters \( (OD_{FEOT}) \) and orifice sizes \( (D_{orifice}) \) are used to investigate in more detail how liquid output rates and residual have changed. Fig. 2 gives a schematic representation of the FEOT-based MHVMN. The gap size between the liquid reservoir and a FEOT, \( L_{LR-FEOT} \) (mm), is given by

\[
L_{LR-FEOT} = \frac{(9.3 - OD_{FEOT})}{2}
\]  

The major operating parameters and ranges are listed in Table 2. The \( L_{LR-FEOT} \) and \( D_{orifice} \) values range from 0.15~0.35 mm and 0.5~2.0 mm, respectively. The effect of nebulizer orientations (upright (0°), horizontal (90°), and inverted (180°)) on output rate and residual is of particular interest. The diameter of orifice on FEOTs is important for solution delivery, especially when the MHVMN is operated in an inverted position. The packing density of the PE wicks used in this study is about 20%. At a rough estimate, the contact areas formed between the mesh-wick structure for strip and rod wicks are 18 and 35 mm², respectively. Besides the passive capillary liquid transport method, a syringe pump (KDS 200, KD Scientific Inc., Holliston, MA, USA) is used to deliver liquid actively to explore the maximum liquid consumption rate of the MHVMN and the effect of liquid feeding rate on the size distribution of generated aerosol particles.
In order to characterize the particle size distributions, the MHVMN is operated in an acrylic chamber. The aerosol flow tube system is shown schematically in Fig 3. The flows in the experiments are generated from compressed air passed through particle filters and a dryer prior to use. The total air flow through the chambers is regulated by a mass flow controller (MFC, Teledyne Hastings Instruments, Hampton, VA, USA). A flow divider valve is used downstream of the MFC to split the total flow into two parts: a dispersion flow and a dilution flow. The generated droplets are first mixed with a dispersion flow of 1.0 L min\(^{-1}\) and then passed through a 10 mCi \(^{241}\)Am neutralizer positioned immediately downstream of the MHVMN. The charge neutralized droplets are then mixed with 159 L min\(^{-1}\) of dilution air to evaporate the solvent and to obtain a stable particle size distribution. The number concentration and size distribution of the aerosol particles are monitored using an aerosol size spectrometer (Welas-3000, Palas GmbH, Karlsruhe, Germany), covering particle size ranging from 0.3 to 40 \(\mu m\) and aerosol number concentration up to \(10^5\) count cm\(^{-3}\). Polystyrene latex spheres (Duke Scientific Corporation, Palo Alto, CA, USA) are used to check the size accuracy of the Welas-3000. Moreover, a Micro Orifice Uniform Deposit Impactor, (MOUDI, model 110, MSP Corp., St. Paul, USA), is employed to measure aerosol mass concentrations and size distributions, covering particle size ranging from 0.056 to 18 \(\mu m\). Silicone-grease-coated aluminum foils (47mm) are used as collection media and are weighed on an analytical balance (Precisa 92SM-202A, Teopal,
Switzerland) before and after each sampling run to determine the mass distribution. Then, the mass median aerodynamic diameter (MMAD) and geometric standard deviation (GSD), defined by the following equations, of aerosol are calculated on a log-probability plot with the cut-off diameter as the abscissa and the cumulative percent less than the cut-off size as the ordinate.

\[
\text{MMAD} = \frac{d_{84\%}}{d_{50\%}}
\]

(3)

\[
\text{GSD} = \frac{d_{84\%}}{d_{50\%}} = \left(\frac{d_{84\%}}{d_{16\%}}\right)^{\frac{1}{2}}
\]

(4)

Other important characteristics of nebulizer performance include output rate and residual. In this study, these characteristics of the MHVMN are determined by gravimetric methods. The MHVMN weight is measured before and after each nebulization experiment, and the nebulization time is also recorded. The output rate and residual (percentage of the retained nebulizer charge at the cessation of nebulization) are calculated according to the equations below. This process is repeated at least three times for each test run.

\[
\text{Output rate} = \frac{\text{Postweight} - \text{Preweight}}{\text{duration of nebulization} \times \text{solution density}}
\]

(5)

\[
\text{Residual} = \frac{\text{Postweight} - 60}{\text{Preweight} - 60} \times 100\%
\]

(6)

The MHVMN is mainly evaluated with deionized water (DI water) and 0.9% sodium chloride solution (NaCl). The 0.9% NaCl solution has a surface tension of 72.5 mN m\(^{-1}\), a density of 1.005 g cm\(^{-3}\), and a viscosity of 1.019 mPa·s (Masoodi and Pillai, 2012).
RESULTS AND DISCUSSION

Effect of \( L_{\text{LR-FEOT}} \) on FEOT-based MHVMN performance

The gap between two coaxial tubes \( (L_{\text{LR-FEOT}}) \) is a critical factor influencing the output rate of the MHVMN, as shown in Fig. 4, with the \( D_{\text{orifice}} \) of 1 mm, and DI water as the test agent. The output rate of the upright MHVMN is the most typical case. As shown in Fig. 4, the output rate increases with increasing \( L_{\text{LR-FEOT}} \) up to 0.225 mm, and then decreases as the \( L_{\text{LR-FEOT}} \) kept increasing, indicating that the gravity gradually dominate the flow process when the \( L_{\text{LR-FEOT}} \) increase. The effect of gravity on liquid delivery in the MHVMN is more obvious when the MHVMN is in inverted position. The output rate increases with increasing \( L_{\text{LR-FEOT}} \), apparently due to more weight and less drag on the liquid as the \( L_{\text{LR-FEOT}} \) widens. As expected, the output rate of the horizontal case is almost unaffected by \( L_{\text{LR-FEOT}} \).

The residual is also an important performance index, especially for expensive medicines. The lower the residual, the better the nebulizer performs. The residual of this MHVMN is less than 10% for all positions, as shown in Fig. 5, with the same \( D_{\text{orifice}} \) (1 mm), and DI water as the test agent. The residuals of both upright and inverted positions decrease slightly with increasing \( L_{\text{LR-FEOT}} \). This is because the inner tube tended to tilt when the gap is too wide, as observed in the laboratory, and thus the capillary transport capability dropped. When the MHVMN is operated in horizontal position, the gap between the two coaxial tubes tend to become non-uniform because
the inner tube would fall and make contact with the inner wall of the outer tube, and the orifice on
top of the inner tube is not on the center of the vibrating mesh. Therefore, the residual is slightly
higher than the others. The residual of the inverted MHVMN is the lowest due to the influence of
gravity. In general, the upright MHVMN with $L_{LR-FEOT}$ of 0.225 mm has the highest output rate of
0.24 mL min$^{-1}$ and a residual of 3.5%. When this design is placed inverted, the output rate is 0.27
mL min$^{-1}$, with a residual of 2.3%.

Effect of $D_{orifice}$ on FEOT-based MHVMN performance

The $D_{orifice}$ has significant influence on the performance in all MHVMN positions. The effect
of orifice size on MHVMN output rate and residual is shown in Fig. 6 and Fig. 7, respectively.
The $L_{LR-FEOT}$ of 0.225 mm is chosen for the highest output rate shown in Fig. 4. For the upright
case, the output rate is only 0.1 mL min$^{-1}$ when the smallest orifice, 0.5 mm, is used, and the
residual is as high as 90%, because the siphon function is frequently stopped by the solution film.
The residual decreases sharply when $D_{orifice}$ increases from 0.5 mm to 0.8 mm, indicating that the
chance of film formation to cover the orifice also decreases abruptly. The output rate reaches the
maximum when $D_{orifice}$ is 1.0 mm, followed by gradual decrease in output rate with increasing
$D_{orifice}$. This is because of less contact areas formed between the mesh-orifice structure for upright
MHVMN with larger $D_{orifice}$. For the inverted MHVMN, the residual is similar to that of the
upright case. Yet, the output rate increases linearly with increasing $D_{\text{orifice}}$. The output rate for a 2.0-mm orifice is $0.72 \text{ mL min}^{-1}$, indicating dominant influence of gravity.

Performance comparison of FEOT-based and wick-based MHVMN

Without considering the issues of cleaning and maintenance, it has been shown in a previous study (Kuo et al., 2019) that porous materials can be used for solution delivery in a VMN. Therefore, two types of PE wicks, strip and rod, are used in this work to deliver the solution to the vibrating mesh plate for further atomization. The PE strip and PE rod have the same packing density (20%), but different contact areas formed between the mesh-wick structure. As shown in Fig. 8, MHVMN with PE rod has higher output rate and higher residual than the one with PE strip due to larger contact area and larger wick volume, respectively. Although the output rates of the two PE wicks used in this work are comparable to that of FEOTs, the FEOT-based MHVMN is superior to the wick-based one for much lower residual. The effect of wick properties on output rate and residual is not within the scope of the present study but merits further exploration.

Effect of solution on the performance of FEOT-based MHVMN

Aerosol therapy may be carried out using an undiluted drug solution or it may require dilution beforehand. Isotonic sodium chloride solution (0.9% NaCl) is widely used for dilution purposes. Thus, two types of solutions, DI water and 0.9% NaCl, are used in this study to examine whether
different solutions would have different output rates and residual. The tests are conducted using MHVMNs with $L_{LR-FEOT}$ of 0.225 and 0.350 mm, respectively. Fig. 9 shows negligible differences in output rate, probably because both solutions have about the same surface tension.

**Effect of liquid feeding rate on the performance of vibrating mesh**

The maximum output of this vibrating mesh is determined using a syringe pump, as shown in Fig. 10. The output concentration increases linearly with increasing 0.9% NaCl feeding rate, and reaches the maximum when the feeding rate is 0.8 mL/min. The maximum number concentration is 19500 count cm$^{-3}$ in the mixing chamber. Aerosol generation terminated when feeding rate exceeds 0.8 mL min$^{-1}$, because the mesh plate is overloaded to change the resonance frequency. Fig. 11 shows the number concentrations and size distributions of aerosol particles, with 0.9% NaCl feeding rates varying from 0.1 to 0.8 mL min$^{-1}$. As can be seen, the concentration increases linearly with feeding rate while the size distribution remained unchanged, with count median diameter (CMD) of 0.65 μm and GSD of 2.3.

**Effect of orientation and aperture size on the performance of FEOT-based MHVMN**

Since the aerosol size spectrometer might have detection efficiency problem for particles too large or too small. Mass concentration and size distribution of the generated droplets after being dried are also validated using the MOUDI. Fig. 12 shows the aerosol mass concentration as
function of aerodynamic diameter for the FEOT-based MHVMN with the best reservoir design 
($L_{LR\text{-FEOT}}$ of 0.225 mm and $D_{orifice}$ of 1.0 mm) operated at different orientations. With 0.9% NaCl
solution and aperture size of 9.5 μm, the generated particle size distributions under three
positions shows the same MMAD of 4.41~4.55 μm and GSD of 2.39~2.93. The aerosol output is
expected to show a log-normal distribution, if the MOUDI is equipped to further separate
aerosols larger than 18 μm. The effect of aperture size on size distribution of nebulized aerosols
is shown in Fig. 13. When smaller aperture size of 6.5 μm is used instead of 9.5 μm, the MMAD
shrinks from 4.5 μm to 3.3 μm. Therefore, the particle size distribution of MHVMN could be
tailored to suit specific requirements by altering the aperture size of the mesh plate.

CONCLUSIONS

A FEOT-based MHVMN that is able to work equally well in any orientation was developed
and fabricated in the present study. Operating parameters including $L_{LR\text{-FEOT}}$, $D_{orifice}$, orientation,
and aperture size were examined to investigate their effects on the performance of MHVMN.
The output rate and residual of the FEOT-based MHVMN was affected not only by $L_{LR\text{-FEOT}}$
and $D_{orifice}$, but also by nebulizer position as well. For horizontal position, the output rate was not
affected by $L_{LR\text{-FEOT}}$; for an inverted MHVMN, the output rate increased with increasing $L_{LR\text{-FEOT}}$
because of the dominating gravity effect. The $L_{LR\text{-FEOT}}$ of 0.225 mm appeared to be the optimal
design because of the output rate and residual were similar for different orientations.
The $D_{orifice}$ had to exceed 0.8 mm to guarantee continuity of nebulization, thus reducing residual. The FEOT-based MHVMN with $D_{orifice}$ of 1.0 mm had the highest output rate and minimum residual at an upright position. The FEOT-based MHVMN was superior to the wick-based one for its lower residual volume.

Furthermore, size distribution of the aerosol particles generated by the FEOT-based MHVMN with the best reservoir design ($L_{LR-FEOT}$ of 0.225 mm and $D_{orifice}$ of 1.0 mm) was not affected either by the nebulizer position. Therefore, the FEOT-based MHVMN is an ideal aerosol delivery device for pure aqueous and normal saline drug solutions, especially for patients lying on bed.

REFERENCES


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Hypertension. Pulmonary circulation 8: 2045894018809084-2045894018809084.


Table 1. Technical information of MHVMN

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<th>Handheld nebulizer</th>
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<td>Battery voltage, V</td>
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<tr>
<td>Power consumption, W</td>
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<tr>
<td>External dimensions, cm</td>
<td>6(L) × 4(W) × 4(H)</td>
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<td>Weight, g</td>
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<table>
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<td>Number of apertures</td>
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Table 2. Operating parameters and ranges

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<td>Upright (0°)</td>
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<tr>
<td>Horizontal (90°)</td>
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<td>Inverted (180°)</td>
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<table>
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<td>Passive</td>
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<tr>
<td>Outer diameter of FEOTs (OD_{FEOT}), mm</td>
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<tr>
<td>Gap between coaxial tubes (L_{LR-FEOT}), mm</td>
<td>0.15 ~ 0.35</td>
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<tr>
<td>Diameter of orifices on FEOT (D_{orifice}), mm</td>
<td>0.5, 0.8, 1.0, 1.4, 2.0</td>
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<tr>
<td>Shape of PE wicks</td>
<td>Strip, Rod</td>
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<tr>
<td>Active</td>
<td></td>
</tr>
<tr>
<td>Syringe pump, mL min⁻¹</td>
<td>0.1 ~ 1.0</td>
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</tbody>
</table>
**Figure Captions**

**Fig. 1.** Prototype of MHVMN: (A) assembly view of vibrating mesh and liquid delivery unit, and (B) three passive liquid delivery materials.

**Fig. 2.** Schematic diagram of FEOT-based MHVMN. Right plot is the blow-up figure of the mesh plate, showing the orifice on top of the inner tube.

**Fig. 3.** Schematic diagram of experimental system set-up.

**Fig. 4.** Effect of $L_{LR-FEOT}$ on output rate of FEOT-based MHVMN in different orientations.

**Fig. 5.** Effect of $L_{LR-FEOT}$ on residual of FEOT-based MHVMN in different orientations.

**Fig. 6.** Effect of $D_{orifice}$ on output rate of FEOT-based MHVMN in different positions.

**Fig. 7.** Effect of $D_{orifice}$ on residual of FEOT-based MHVMN in different positions.

**Fig. 8.** Performance comparison of FEOT-based and wick-based MHVMNs.

**Fig. 9.** Effect of solution on performance of FEOT-based MHVMN with different $L_{LR-FEOT}$.

**Fig. 10.** Output aerosol concentration as function of solution feeding rate.

**Fig. 11.** Effect of feeding rate on particle size distribution.

**Fig. 12.** Aerosol mass concentration as function of aerodynamic diameter for FEOT-based MHVMN with the best reservoir design operated at different orientations.

**Fig. 13.** Effect of aperture size on particle size distribution of FEOT-based MHVMN.
Fig. 1.

(A) Fixing cap
Vibrating mesh plate
Solution reservoir ($\varnothing 9.3 \times 35 \text{ mm}^2$)

(B) Strip PE wick
Rod PE wick

Flat end orifice tube ($FEOT$)

Fig. 2.
Fig. 3.

Fig. 4.
Fig. 5.

$D_{\text{orifice}} = 1 \text{ mm}$
Solution: DI water

Orientation
- Horizontal (90°)
- Upright (0°)
- Inverted (180°)

$L_{LR-\text{FEOT}}$, mm
Residual, %

Fig. 6.

$L_{LR-\text{FEOT}} = 0.225 \text{ mm}$
Solution: DI water

Orientation
- Inverted (180°)
- Upright (0°)

Output rate, mL min$^{-1}$

$D_{\text{orifice}}$, mm
Fig. 7. $L_{LR-FEOT} = 0.225 \text{ mm}$
Solution: DI water

Fig. 8.
Fig. 9. 

Fig. 10.
Fig. 11.

- Solution: 0.9% NaCl
- Feeding rate, mL min⁻¹:
  - 0.1: 0.66, 2.26
  - 0.2: 0.64, 2.21
  - 0.3: 0.65, 2.38
  - 0.4: 0.65, 2.42
  - 0.6: 0.66, 2.33
  - 0.8: 0.66, 2.28

Fig. 12.

- Orientation: upright (0°)
  - MMAD: 4.05 μm
  - GSD: 2.39
- Orientation: horizontal (90°)
  - MMAD: 4.47 μm
  - GSD: 2.93
- Orientation: inverted (180°)
  - MMAD: 4.45 μm
  - GSD: 2.84
Orientation: upright (0°)
Solution: 0.9% NaCl

Aperture: 6.5 μm
-- MMAD: 3.32 μm
-- GSD: 2.14

Aperture: 9.5 μm
-- MMAD: 4.52 μm
-- GSD: 2.41

Percent less than indicated size

Aerodynamic diameter, μm

Fig. 13.