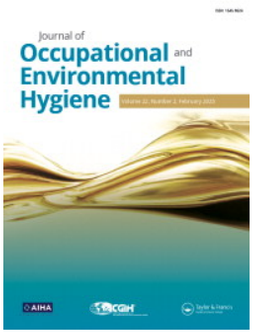

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SHORT REPORT

A cough simulator constructed from off-the-shelf and 3D-printed components

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ABSTRACT

The development of low-cost research equipment is crucial for enhancing accessibility in scientific research, particularly in the field of respiratory disease transmission. This study presents a novel, customizable cough simulator designed for ad-hoc studies that require precise control over ejection velocity and aerosol size. Constructed from off-the-shelf parts and 3D-printed components, this programmable, piston-driven simulator offers an affordable solution for researchers. Its performance has been validated, demonstrating suitability for evaluating fluid flow and monitoring ejected particles that correspond to the velocities of mouth breathing and coughing. Potential applications for this device include assessments of aerosol ventilation, disinfection, and the efficacy of personal protective equipment, all of which contribute to advancing scientific understanding and public health outcomes.

KEYWORDS

Aerosol; human; particles; piston; respirator; transmission

Introduction

The transmission of respiratory diseases is a significant public health concern. While extensive research has been conducted on this topic, there remains a need for effective tools to study the dynamics of respiratory ejecta. This study introduces a novel, low-cost, and customizable cough simulator specifically designed for ad-hoc studies requiring precise control over ejection velocity. Unlike previous simulators, which may be prohibitively expensive or limited in functionality, our cough simulator is constructed from readily available off-the-shelf components and 3D-printed parts, making it accessible to a broader range of researchers. This affordability facilitates exploratory research and empowers students and emerging scientists to gain hands-on experience in respiratory disease research. In addition to its cost-effectiveness, the cough simulator offers programmable, piston-driven flow capabilities that allow for the simulation of various respiratory events with high repeatability. This feature is particularly advantageous for studies evaluating aerosol ventilation, disinfection, and the efficacy of personal protective equipment. By providing a practical solution that combines affordability with advanced functionality, this cough simulator represents a significant advancement in the tools available for studying respiratory disease transmission.

Human emissions depend upon the individual's activity, such as breathing, talking, singing, coughing, and sneezing (Stadnytskyi et al. 2021). Talking produces a range of particle sizes, which average about one micron, and their concentration increases with speaking intensity (Asadi et al. 2019). Coughing and sneezing produce the most significant concentrations of human emissions (Dhand and Li 2020). These violent expiratory events project a potentially infectious mixture of respiratory fluids into a multiphase turbulent buoyant cloud (Bourouiba et al. 2014). Small droplets and particles remain suspended in the air (Leung 2021). Large cohesive droplets containing the most inertial energy travel far and drop quickly onto surfaces where infectious material can last for hours or days as a fomite, depending on the specific pathogen and ambient conditions.

Cough simulators mimic human subjects to study disease transmission. These apparatuses are safer than testing with human subjects—especially during a pandemic when human subjects may transmit diseases. Simulators also provide repeatable emissions, which

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helps to reduce variability between human subjects and even within the same subject. Furthermore, research using simulators can help address safety and privacy concerns typically associated with studies involving human subjects. However, simulators do not mimic all aspects observed in human test subjects.

The human respiratory tract has physiological mechanisms that produce emissions with specific particle size, composition, and momentum. The primary mechanism is breathing. With each breath, the body emits warm, humid air with particles created by liquid film rupture, also known as bronchial fluid film burst (Haslbeck et al. 2010; Morawska et al. 2022). The particles emitted during breathing vary in size, ranging from tens of nanometers up to ten microns, with the peak concentration found to be around one micron in size. Speech produced additional particles in modes near 3.5 and 5 μm (Morawska et al. 2009). Additional oral emission modes during speaking were observed between 8 and 13 μm and between 90 and 200 μm (Johnson et al. 2011) with all previous particle modes underlying cough emissions (Pöhlker et al. 2023). Lastly, sneezes produce the largest emissions. Sneezes are the most energetic exhalation and produce the largest droplets with long bands of tissue and sheets (Scharfman et al. 2016).

Considering the multiple size ranges or "size modes," cough simulators typically only represent one or two size modes and do not replicate bands of tissue and sheets. Some simulators use a fluid consisting of a mixture of distilled water and glycerin, illuminated by a laser light sheet or by Schlieren imaging in recent studies (Staymates 2020; Verma et al. 2020). These studies also use a mannequin to show differences in airflow around and through face-worn products that provide a barrier between the nose and mouth of the wearer and the fluid mixture. Another approach to simulating a single mode of human ejecta is to use polydisperse NaCl solution in combination with size-resolved particle counting instruments (Edwards et al. 2020). Other studies have aerosolized live viruses and compared the size distribution of aerosol generated from an airbrush to an aerosol generated with a micropump nebulizer (Lindsley et al. 2013). The cough simulator herein was used to evaluate a local ventilation system to mitigate retail store workers' exposure to airborne particles (Lee et al. 2023). A bimodal cough simulator uses an atomized aerosol stream in which a second fluid is injected (Zhang et al. 2017). These types of advances help to simulate essential properties of human ejecta.

Simulators have been evaluated for the physiological relevance of droplet size, 3D spread, velocity, and force (Patel et al. 2020). However, because of the dissimilarities between results, Patel et al. recommend caution when comparing conclusions across simulator studies. Compared to the low-cost cough simulators evaluated by Patel et al. the cough simulator herein provides controllable and relevant velocities.

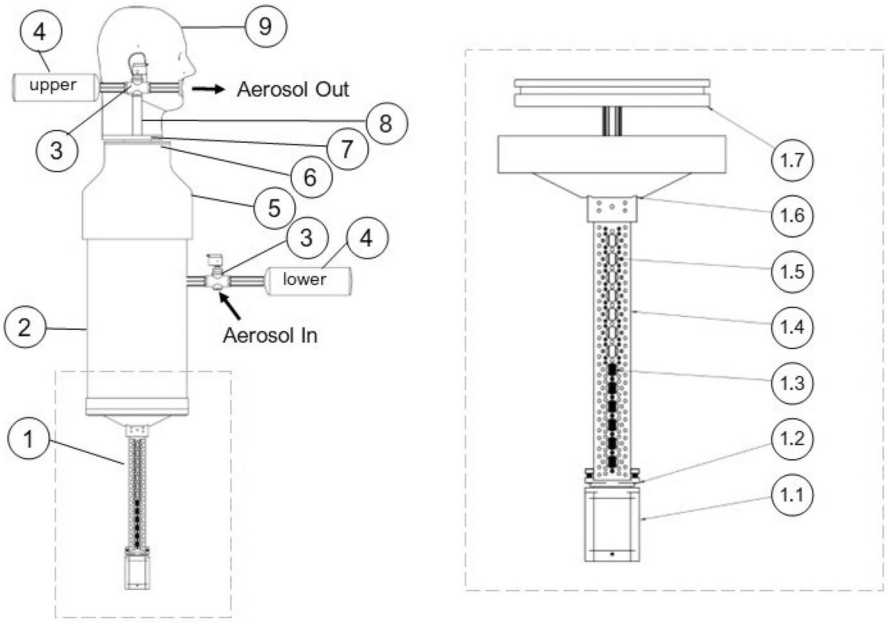
This paper presents a low-cost cough simulator built from off-the-shelf and 3D-printed parts for a few hundred dollars. It offers programmable control of exhalations, though this study used high-end equipment; cheaper options like a basic spirometer and particle counter could be substituted. Affordable research tools are crucial for studying respiratory disease transmission, enabling more researchers to contribute. The simulator allows precise control of ejection velocity and aerosol size using a piston-driven system. Results show it's an affordable option for studying fluid flow and particles linked to coughing and mouth breathing.

Methods

The motion was generated through an Actobotics X-Rail liner actuator kit SKU: 637213 (7.4″ stroke, ServoCity.com, Winfield, KS). The actuator kit was powered with a National Electrical Manufacturers Association (NEMA) 57 Integrate Stepper Servo Motor (Model iHSS57-36-20, Just Motion Control Electro-Mechanics Co., Ltd, Shenzhen, China) and attached by a NEMA 57 motor mount. An 8-mm Lead Screw, 10.50″ Channel, and 10.50″ X-Rail were included in the Actobotics X-Rail linear actuator kit and are itemized in Figure 1.

The part named Rail-to-8″ polyvinyl chloride (PVC) (Figure 1, Part 1.6) that connects the rail to the PVC pipe and the part named Baseplate-to-Piston (Part 1.7.a) that connects the Baseplate to the Piston were 3-D printed with acrylonitrile butadiene styrene (ABS) filament. Different printer settings and different-sized O-rings were evaluated to find a snug fit. To provide an airtight seal, the 8″ PVC pipe (Part 2) was glued on each side (i.e., to Part 5 and Part 1.6) using PVC glue and primer.

The motorized, 3-way ball valve (Part 3) has three ½″ female national pipe thread (FNPT) ports (HSH-Flow, Jiangsu and Zhejiang Provinces, China). The 3-way valve was attached to a high-efficiency particulate air (HEPA) filter with a ½″ make national pipe thread (MNPT) pipe (Part 4). One of the 3-way valve and HEPA filter combinations was attached inside the National Institute for Occupational Safety and Health (NIOSH) head form (Part 9), and another was



#	Part name
1	Piston Assembly
2	8" PVC pipe
3	3-way valve
4	HEPA filter
5	8" to 4" PVC
6	4" to 2" PVC
7	2" to 1/2" PVC
8	1/2" PVC pipe
9	NIOSH Headform

#	Part name	Model	Manufacturer
1.1	NEMA 57 Stepper	IHSS57-36-20	Toauto
1.2	NEMA 57 Mount	SKU: 555160	ServoCity
1.3	8mm Lead Screw	SKU: 545315	ServoCity
1.4	10.50" Channel	SKU: 585452	ServoCity
1.5	10.50" X-Rail	SKU: 565042	ServoCity
1.6	Rail-to-8" PVC	3D Printed	
1.7a	Baseplate-to-Piston	3D Printed	
1.7b	Round Baseplate	SKU: 585438	ServoCity
1.7c	Buna-N O-Ring	Dash 441	Mr o-ring

Figure 1. Schematic of the cough simulator components.

attached to the 8″ PVC pipe body (Part 2)—referred to as the "upper" and "lower" valve, respectively. In addition to the HEPA filter, the upper valve was also attached to a pipe to ½″ PVC pipe "throat" on one side and to a ½″ PVC pipe extending out from the head form and attached to a laboratory nozzle "mouth" with an 8.8-mm-diameter opening. The lower valve was attached to the 8″ PVC body (Part 8) on one side and to a tube providing the generated aerosol. Schedule 40 PVC pipe fittings were used to reduce the 8″ pipe to ½″ pipe (Part 8) in the following steps: 8″ to 4″ (Part 5), 4″ to 2″ (Part 6), and 2″ to 1/2″ PVC (Part 7) (see Figure 2).

The NEMA 57 motor received power from a 350-watt switching power supply (Model LRS-350-48, Mean Well USA, Inc., Fremont, CA). Input pulses to control the motor movement were generated by a Raspberry Pi 4 Model B single-board computer (Raspberry Pi Foundation, Cambridge, UK). The computer also received contact from electrical limit switches at the beginning and end stroke of the piston. The beginning stroke limit switch sets the "zero" start position. The end stroke limit switch provided a safety shutoff to turn off movement beyond the maximum length of the piston. The computer low-level DC outputs provided signals to 24 VDC relay switches that opened and closed the 3-way valve described previously.

Piston movement

The speed at which a volume of fluid moves is called the volumetric flow rate and is defined by the following equation:

Figure 2. Photo of the cough simulator.

$$Q_1 = v_1 * A_1 \tag{1}$$

where:

Q1 is the volumetric flow rate in m^3/sec

v1 is the mean velocity in m/sec

A1 is the cross-sectional vector area in m^2

The volume of air leaving the 8.8-mm orifice extending out from the head form—which will be referred to as the "mouth"—is equal to the volume of air displaced by the piston (i.e., Q1 = Q2 or v1*A1 = v2 * A2). The volume change is proportional to the ratio of areas (see Equation 2).

$$v_1 * \frac{A_1}{A_2} = v_2 \tag{2}$$

The actual inner diameter of the mouth was 8.8 mm (0.0088 m) when measured with precision calipers, and the actual inner diameter of the 8″ PVC pipe was 201.7 mm (0.2017 m). The ratio of the areas was determined with Equation 3.

$$\frac{A_1}{A_2} = \frac{\pi\left(\frac{0.0088}{2}\right)^2}{\pi\left(\frac{0.2017}{2}\right)^2} = 0.0019035 \tag{3}$$

v1*0.0019035 = v2

The motor step angle of 1.8 degrees produces one revolution per 200 steps. With a dip switch setting to 1/4 step, the motor produces one revolution per 800 steps. The lead screw has a 2-mm pitch with four starts that produce 8 mm of forward movement per revolution. Thus, 800 steps provide 8 mm of forward movement, or 1 m is 100,000 steps.

Python 3.7 was used to write software that controls the motors' and relays' timing and movements. The software was operated through a cross-platform GUI toolkit (PyQt5). Clicking the Start button on the computer screen initiated a sequence of steps. The first step expelled air from the 8″ PVC pipe through a HEPA filter. The upper and lower valves were set as listed in Table 1. The piston moved to the extension limit near the top of the 8″ PVC pipe. The second step was to fill the 8″ PVC pipe with aerosol until the aerosol was seen exiting the mouth and then move the piston to the retraction limit near the bottom of the 8″ PVC pipe. The retraction speed matched the aerosol's flow rate so that the 8″ PVC pipe filled without drawing air through the HEPA filter. The third step was to eject aerosol out of the mouth for a cough or breathing sequence.

For exhalations, the piston velocity was set to 0.00247 m/sec for simulated breath exhalation and 0.02665 m/sec for simulated cough exhalation. However, the piston did not provide a complete seal. The volumetric flow rate at the mouth was increased to 0.00675 m/sec and 0.08700 m/sec to account for leakage around the piston and loss in the system. The increase was determined empirically to achieve the desired exit velocities of 5.0 lpm and 50 lpm, respectively.

Only exhalations were evaluated in this study. The inhalations were used to fill the piston with aerosols. Researchers tested various dwell times and retraction rates. Dwell time refers to the time required for aerosols to fill the throat area and visibly exit the mouth.

Table 1. Sequence of valve and piston movement.

Step	Purpose	Upper Valve Setting	Lower Valve Setting	Piston Movement
1	Evacuate air out from the 8″ PVC pipe through a HEPA filter	Throat to HEPA filter; closed to mouth	Aerosol to HEPA filter; closed to 8″ pipe	Move the piston to the extension limit
2	Fill the 8″ PVC pipe with aerosol	Throat to HEPA filter; closed to mouth	Aerosol to 8″ pipe; closed to HEPA filter	Move the piston to the retraction limit
3	Eject aerosol out the mouth	Throat to mouth; closed to HEPA filter	Aerosol to HEPA filter; closed to 8″ pipe	Move the piston to a cough or breath sequence

Retraction rate refers to the speed at which the inhalation fills the piston with the generated aerosols. The inhalation flow rate was adjusted to match the flow rate of the aerosol generation, ensuring that the piston was filled without drawing ambient air through the mouth.

The piston movements were linear for this study. Rapid starting and stopping help evaluate any system shocks and leaks. Different exhalation intensities and durations could also be programmed with values to match a normal tidal breathing exhalation (Scott and Kaur 2020).

Results

The volumetric flow rate for simulated breathing and coughing is plotted *vs.* time in Figure 3. The flowmeter reported at a 1-MHz frequency (i.e., 0.000001 readings per second). The total volume of expelled air was 0.445 ± 0.0003 L for breathing and 0.087 ± 0.0029 for coughing. The breathing and coughing duration were 5.5 ± 0.002 and 0.101 ± 0.0023 sec, respectively.

The piston movement associated with breathing showed a well-defined beginning and end. The peak volumetric flow rate occurred near the onset of the piston movement and then remained consistent throughout the event. The quick piston movement associated with coughing increased the flow rate for approximately 0.05 sec, maintained a peak for about 0.05 sec, and then declined over the next 0.05 sec. The breathing and coughing events were highly repeatable concerning airflow magnitude and time, as evidenced by the small standard deviations associated with repeated volumetric flow measurements.

Glass-sphere particles (a mixture of $d50 = 4$ and 10 µm, Cospheric, Santa Barbara, CA) were generated by a fluidized bed aerosol generator (Model 3400, TSI Inc.) and passed through an aerosol neutralizer (Model 3034, TSI Inc.). Polydisperse sodium chloride (NaCl) particles were dissolved in ultrapure (type 1) water and generated using a 24-jet Collison nebulizer (CH Technologies Inc., Westwood, NJ, USA) through a diffusion dryer (Model 3062, TSI Inc.). Particle size distributions of the polydisperse glass spheres and NaCl are shown in Figure 4. The peak concentration was nearly 7×10^3 counts/cm^3 for NaCl and 7×10^7 counts/cm^3 for glass spheres. The aerosol concentrations are many times greater than those emitted by humans. The unnaturally high concentrations were used to provide ample particles for the particle counting instruments and mask the room's background particle concentration.

The average count median aerodynamic diameter (CMAD) during mouth breathing (1.79 ± 0.05 µm) was slightly greater than that during coughing (1.46 ± 0.03 µm). One likely explanation of why the breath particles were larger than the cough particles would be due to the difference in speed through the upper valve. The rapid speed of the cough within a 90° bend in the upper valve likely caused large particles to impact on the perpendicular surface. This aerosol impaction would disproportionally favor large particles and allow smaller particles with less inertial energy to pass through without impaction. This impaction surface also explains why the median aerodynamic diameters of the glass sphere particle measurements herein were less than those indicated by the manufacturer ($d_{50} = 3$–6 µm). Subsequent design

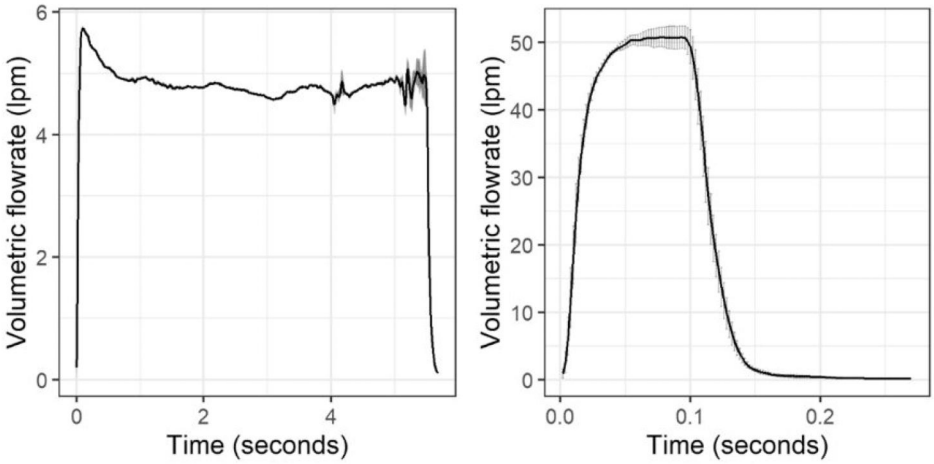


Figure 3. Volumetric flow rate at the aerosol outlet of the cough simulator during mouth breathing (left) and coughing (right). Mean and standard error ($n = 9$ and $n = 12$, respectively).

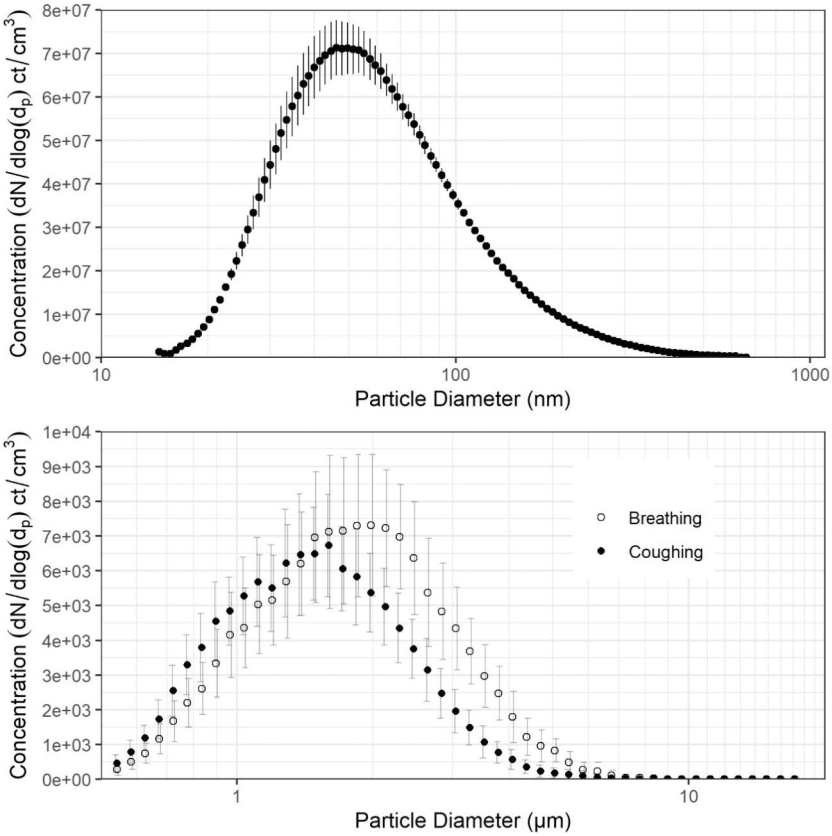


Figure 4. Particle count of mass median aerodynamic diameter of sodium chloride in nm (upper) and polydisperse glass sphere in μm (lower). Mean and standard error ($n = 5$).

improvements could include physiologically relevant modeling of the human respiratory system.

Discussion

The study presented a novel, low-cost cough simulator constructed from off-the-shelf and 3D-printed components, designed to facilitate research on respiratory disease transmission by allowing precise control over ejection velocity and aerosol size. The simulator's performance was validated through a series of measurements that demonstrated its ability to produce highly repeatable volumetric flow rates corresponding to human mouth breathing and coughing. Specifically, the total volume of expelled air was measured, showing values of 0.445 L for breathing and 0.087 L for coughing, with associated durations of 5.5 and 0.101 sec, respectively. The study reported that both breathing and coughing events exhibited well-defined beginning and endpoints, with the peak volumetric flow rates occurring near the onset of piston movement and remaining consistent throughout the events. The repeatability of these measurements was evidenced by small standard deviations associated with repeated volumetric flow measurements, indicating that the simulator could reliably replicate the dynamics of human respiratory events. Results indicated that the simulator produced highly repeatable volumetric flow rates, with specific measurements showing a total volume of expelled air of 0.445 L for breathing and 0.087 L for coughing, alongside distinct particle size distributions for each respiratory event.

Air leakage was observed at the piston O-ring. A procedure for measuring air flow rates at the mouth—such as the one described in the Methods section—should be used to account for air leakage in the system. The leakage could increase the ambient room particle concentration. This increase could increase the concentration of particles measured; therefore, leakage control, such as an internal device ventilation system or a more efficient piston, should be considered when using the device. Leakage control, such as by room ventilation, a more efficient piston, or a secondary containment around the simulator, should be considered when using the device.

The particle size distribution was different between the mouth breathing and coughing settings and

different from the manufacturer's specification. As stated previously, it is reasonable that the 90-degree bend in the upper valve reduces the concentration of larger particles according to the principle of inertial impaction. Eddie impaction at low velocities is also a likely mechanism for the deposition of particles larger than one micron that experience turbulent airflow (Im and Ahluwalia 1989).

The higher aerosol concentrations generated by the cough simulator, which are over a million times greater than actual human breathing emissions for some particle sizes, significantly impact the validity and applicability of the simulator. This artificially elevated concentration is intended to ensure sufficient particles for accurate counting and to mask ambient aerosol concentrations, which can obscure the small amounts typically emitted by humans.

While this approach allows for precise measurements and broad distribution spectrum analysis, it raises concerns regarding the biological validity of the results, particularly when considering the concentration of potentially infectious particles that would be emitted by a human. The simulator's outputs may not accurately reflect real-world scenarios where human-generated aerosol concentrations. Consequently, findings derived from the simulator should be interpreted with caution, particularly when extrapolating to human behavior or public health implications.

The simulator is best suited for studies focused on relative measurements, such as the percentage reduction of small particles in controlled environments, rather than for direct comparisons to human aerosol emissions. Therefore, while the simulator provides valuable insights into aerosol dynamics and ventilation efficacy, its results must be contextualized within the limitations of its design, particularly regarding aerosol concentration.

The sizes of the generated aerosol (20 nm to 4 μm) were comparable to the range of sizes emitted from human breathing (Pöhlker et al. 2023). However, the percentage of particles at a particular size (i.e., the size distribution) of the generated aerosol did not match human breathing.

Considering that the particle size range was comparable to human breathing, but the particle concentration and size distribution were not, only relative measurements are appropriate.

The percentage reduction in particles from one location to another is an acceptable relative measurement. For example, if 1000 one-micron particles were counted at point A and 500 one-micron particles were counted at point B, the percentage reduction of particles is 500/1000 or 50%. Additional one-micron particles released at point A and following the same path would also diminish by 50%, although the concentration of particles would be different.

Coughing includes the underlying breath-related bronchial particle formation but also involves an abrupt parting of the vocal cords and sudden expulsion of breath that generates additional distributions of large droplets (Pöhlker et al. 2023). Single-mode cough simulators such as the device evaluated herein do not generate these droplets. Single-mode cough simulators are suited for simulating breathing and the breath component of cough but not the projection of large droplets.

Considering this limitation, there are better choices than single-mode cough simulators for studies on large droplets, such as when evaluating the gravitational deposition of droplets onto surfaces. However, evaluating air movement, air mixing, and percentage reduction of small particles that remain aloft for extended periods is a good fit for single-mode cough simulators. Evaluation of ventilation systems and their ability to prevent the spread of cough aerosol is an excellent example of a study of air movement and mixing.

Conclusion

The cough simulator developed in this study effectively simulates velocities comparable to those associated with human mouth breathing and coughing, demonstrating unique and highly repeatable volumetric flow rates. Air leakage was observed at the piston O-ring, necessitating a procedure for measuring airflow rates at the mouth to account for this leakage. The particle size distributions of sodium chloride and glass spheres differed between simulated breathing and coughing, with the simulator being most applicable for evaluating air movement and mixing, particularly in assessing the percentage reduction of small aerosols by ventilation systems.

After considering the particular research question, the cough simulator evaluated herein can be a suitable low-cost option for a single-mode cough simulator to evaluate the percentage reduction of small aerosols.

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Disclaimer

The findings and conclusions in this report are those of the authors and do not necessarily represent the official position of the National Institute for Occupational Safety and Health, Centers for Disease Control and Prevention. Mention of any company or product does not constitute endorsement by the National Institute for Occupational Safety and Health, Centers for Disease Control and Prevention.

Disclosure statement

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Data availability statement

Data will be available upon request.

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