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Optimized mechanism for fast removal of infectious pathogen-laden aerosols in the negative-pressure unit

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HIGHLIGHTS

• Universal virus aerosol removal mechanisms in the clinic unit are suggested using the numerical simulation.
• 12 ventilation types are evaluated in terms of the aerosol removal performance.
• The key strategy is to establish the flow to prevent the immediate dispersion of the aerosols.

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ABSTRACT

It has been frequently emphasized that highly contagious respiratory disease pathogens (such as SARS-CoV-2) are transmitted to the other hosts in the form of micro-sized aerosols (< 5 μm) in the air without physical contacts. Hospital environments such as negative-pressure unit are considered being consistently exposed to pathogens, so it is essential to quickly discharge them through the effective ventilation system. To achieve that, in the present study, we propose the optimized ventilation mechanism and design for the fastest removal of pathogen-laden aerosol using numerical simulations. We quantitatively evaluated the aerosol removal performance of various ventilation configurations (combinations of air exhaust and supply ducts), and found that the key mechanism is to form the coherent (preferentially upward) airflow structure to surround the respiratory flow containing the aerosol cluster. We believe that the present findings will play a critical role in developing the high-efficiency negative-pressure facility irrespective of its size and environments.

1. Introduction

Since the outbreak of COVID-19 in 2019, our everyday life has been affected significantly, and the damaging effects persist owing to the viral mutations and rapid epidemic rate, even after the worldwide vaccination started in 2021. Infectious respiratory diseases, such as influenza, severe acute respiratory syndrome (SARS), Middle East respiratory syndrome (MERS), and now SARS-CoV-2, have been prevalent for past decades, and each incident has thrown scientific challenges of different issues. What we have learned in common, while coping with the infectious diseases based on RNA virus mutating rapidly, is the fundamental importance of controlling the physical transmission based on the fluid mechanics as well as the medical counteraction (Bourouiba, 2021). Many previous studies have claimed that highly contagious respiratory disease pathogens are transmitted to the other hosts without physical contacts, in the form of a spray of droplets and several micron-sized aerosols in the air (Fennelly, 2020; Morawska and Milton, 2020; Yang

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It has also been experimentally observed that the virus-laden aerosols remain infectious in the air for at least three hours (Lewis, 2020). The airborne transmission is more prominent in indoor spaces than in outdoors; the average positivity rate to SARS-CoV-2 of air samples was more than 30% higher in indoors compared to the outdoor sites (Dinoi et al., 2022). In particular, Dinoi et al. (2022) explained that the average concentration of SARS-CoV-2 RNA traces in air samples from the hospital environments was twice as high as that from other indoor environments.

Since the outbreak of SARS-CoV-2, to mitigate the rapid airborne transmission, the public health officials and government organizations have strongly emphasized the use of face masks as a personal hygiene (Fischer et al., 2020; Repici et al., 2020; Kang et al., 2021), limitation of the occupancy number in a meeting group (Jones et al., 2020), and the social distancing (Balachandar et al., 2020; Chen et al., 2020). In Italy, it was reported that the SARS-CoV-2 RNA concentration of air samples measured in various indoor communities (such as the train station, food market, shopping center, and pharmacy) was significantly reduced by the restrictive policies (Conte et al., 2022). Nevertheless, multiple community infections have been frequently initiated in indoor spaces such as restaurants and workplaces (Lu et al., 2020; Park et al., 2020; Liu et al., 2021). The transmission of SARS-CoV-2 among 3 family groups in a restaurant in China and the widespread (more than 40% of the house) infection of the virus in an office in South Korea prove that the pathogen in the aerosol under a poorly-ventilated indoor environment can easily infect the hosts across the wide area (Lu et al., 2020; Park et al., 2020). Among the various indoor environments, the hospitals where the infected patients should take medical treatments and disease diagnostic tests are considered “super spread” of the virus or “origin” of the massive airborne transmission (Correia et al., 2020; Hadei et al., 2020; Liu et al., 2020; Weiland et al., 2021). Even in the same hospital building, the public areas such as the medical staff compartment and toilets had a much higher concentration compared to ICU, CCU, general ventilated patient rooms (Hadei et al., 2020; Liu et al., 2020). In addition, the infectious disease diagnosis room, where infected people would take off their masks and generate virus-laden aerosols, is a place with a high probability of airborne transmission to other hosts. Thus, it is crucial to assess the localized effects depending on the ventilation type (negative, positive, and neutral) and compartment design during the aerosol-generating processes in treating and screening the patients with infections (Weiland et al., 2021).

In the pandemic situation like nowadays, on the other hand, it is required to install a sufficient number of ventilated infectious disease screening/test/clinic compartments because the quick diagnostic tests play a significant role in reducing the transmission of infectious diseases (Black et al., 2020; Studdert and Hall, 2020; Subramanian et al., 2021). While the large facility for the mass testing and group quarantine have been proved to be very effective in isolating the widespread of airborne disease (Black et al., 2020; Studdert and Hall, 2020), it is also increasingly required to operate small-scaled mobile negative-pressure units that can be easily deployed and adapted to different environments flexibly. It is obvious that such areas must be thoroughly sanitized and ventilated between the entrance of each patient (pathogens in the room should be removed completely) (Lynch and Goring, 2020; Sodiq et al., 2021). The pathogen-laden aerosols contain infectious viral particles that survive for three hours in the unventilated environment (Somsen et al., 2020). Whether it is for retrofitting the poorly-ventilated areas or for developing new-design of portable negative-pressure unit, it is crucial to investigate the ventilation mechanism optimized for various compartment design and airflow structures in it.

To understand the ventilation mechanism, previous studies have observed the airflows and contaminants dispersion patterns under the various ventilation configurations in indoor community environments (Abuhagazy et al., 2020; Mesgarpour et al., 2021; Mirzaie et al., 2021) and hospital environments (Qian and Li, 2010; Wang et al., 2021). Some studies also assessed the exhaust of the SARS-CoV-2 virus-laden aerosol to outdoor (Somsen et al., 2020; Mathai et al., 2021; Weiland et al., 2021). It is generally more effective to apply the negative pressure in the hospital environments to exhaust the aerosols rapidly (Weiland et al., 2021). In addition, it was shown that the reduction of airflow mixing and formation of a localized flow in the unit are helpful in enhancing the ventilation efficiency such that the aerosol does not spread throughout the room (Dietz et al., 2020). In terms of the practical application, however, the indoor air ventilation manual (for time allocation and procedure) for test/clinic room quarantine has not been established based on the results of fluid mechanics studies; it is further necessary to investigate the aerosol dispersion pattern corresponding to each step in the procedure taken before and after the personnel (patient) uses the room.

For past decades, researchers have discussed the application of displacement or mixing type ventilation depending on the purpose of the compartment or targets to be controlled to improve the indoor air quality (Qian and Li, 2010; Bhagat et al., 2020; Sodiq et al., 2021; Yang et al., 2021). Also, the ventilation performance in minimized indoor space models were experimentally evaluated (Cooper and Linden, 1996; Linden, 1999; Mingotti and Woods, 2015). For the mixing-type ventilation (commonly applied to air conditioning), the air supply and exhaust are installed at the same height and the temperature in the room is maintained uniformly (Linden, 1999). With the displacement type, on the other hand, the air supply and exhaust are located at the bottom and top of the room, respectively. In this configuration, the indoor airflow goes upward dragging the exhaled aerosols. If there is a source of heat
and respiratory flow (e.g., patient), the upward thermal plume causes an air density stratification (Linden et al., 1990; Linden, 1999; Mingotti and Woods, 2015; Bhagat et al., 2020), i.e., a warm zone above a cool zone, and the contaminants are locked up in the warm zone. Therefore, the displacement ventilation using the ‘lock-up effect’ has been preferred to prevent the spread of pathogens (Qian and Li, 2010; Mingotti and Woods, 2015; Bhagat et al., 2020; Sodiq et al., 2021). Early studies developed a mathematical model for the interface height between zones depending on the extracted mass flow rate and characteristics of buoyant plume (Linden et al., 1990; Cooper and Linden, 1996). Although the interface height and ventilation efficiency increase with ventilation flow rate according to the model, it was recently shown that the interface height becomes insensitive to the flow rate when the ventilation is too strong (Yang et al., 2021). In this case, kinetic energy of the ventilation flow is larger than potential energy due to the stable stratification, which breaks the interface and directly leaves though the outlet; removal of pathogens from the upper layer become difficult. In addition, the energy loss (degradation of removal efficiency) caused by the friction and blockage associated with the room design, number of occupants, and location of furniture should be considered.

In the present study, using numerical simulation, we suggest the optimized ventilation mechanism and associated negative-pressure system, which is further suited for the portable clinic/test room (functioning as a virus sample collection booth) environment. To eliminate the indoor pathogen-laden aerosols quickly such that the personnel is not exposed to the remaining virus from the previous occupant, we are to focus on the following questions: (i) How do airflows change according to various combinations of ventilation (the air supply and exhaust) locations and room interior configuration? (ii) How can we quantify the capability to discharge the pathogens according to the airflows? In general, the diagnostic works in a pandemic situation are required to be done in the closed biological safety cabinets satisfying the Biosafety Level 2 (BSL-2) (WHO, 2020). Therefore, as illustrated in the reference configuration (Fig. 1), the targeted compartment in the present study is combined of the separate doctor and patient rooms, which maintains a negative (−2.5 Pa) and positive (5.0 Pa) pressure, respectively, to prevent the unwanted inflow of aerosols. The curved ceiling design is adopted to guarantee that the suggested compartment can be readily extended to various outdoor environments (e.g., portable in a vehicle or deployed as a tent/booth) where the risk of infection during the medical activities is known to be relatively less (CDC, 2019). Meanwhile, it was pointed out that it takes quite long to remove 99% of the airborne contaminants from the conventional closed (indoor) chamber; it takes 23 min for removing 99% aerosols with the ventilation requirements of US standards (minimum 12 ACH) (CDC, 2003). It was also measured to take 38 min to remove the virus-laden aerosols in the hospital environments with −2 Pa negative pressure applied (Weiland et al., 2021). In contrast to the conventional design of a closed chamber (WHO, 2020), we conceive the idea of installing a connecting duct between the positive (doctor room) and negative (patient room) pressure room to achieve a quick aerosol discharge. Here, it was hypothesized that the duct would allow the flow of the positive-pressure chamber to enter the negative-pressure unit, accelerating the aerosol discharge from the negative-pressure unit while preventing the aerosol leakage back to the positive-pressure unit. Thus, we will focus on the optimization of the connecting duct configuration (position and shape) for the maximum aerosol removal performance.

This paper is organized as follows. In Section 2, we explain the numerical methodology including the physical simulation geometry (ventilation scenarios) and numerical models. In Section 3, we first analyze the interaction between the respiratory and ventilation flows based on the flow velocity distribution and subsequent aerosol behaviors. The performance of tested ventilation configurations resulting from the flow-particle interaction is evaluated according to the patient’s movement. Following that, we discuss the most critical factor of aerosol exhaust performance and the impact of the door opening/closing. Finally, in Section 4, we summarize with the innovative suggestion for the novel ventilation and insights on the pathogen-laden aerosol dynamics learned from the knowledge of multiphase flow.

2. Problem definition and numerical methodology

2.1. Physical model and simulated scenarios

The present portable infectious disease clinic facility was designed as negative and positive pressure units for the patient and doctor, respectively. This portable facility is a prefabricated structure designed to maximize its mobility in a pandemic situation to be dispatched quickly to different places and used to diagnose many patients (including people with less mobility). Thus, it is intended to be used not only in indoors

Fig. 1. Schematic of the infectious disease treatment (diagnostic testing) unit with a patient. The inset at the top-right represents the breath flow rate [L/s] profile along the time. The other inset (bottom-right) shows the breathing patient and virus-laden aerosols (color denotes the aerosol diameter).
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Table 1
Summary of the ventilation configurations with different combinations of the connecting duct, air exhaust locations, and shapes. The configurations of D1-D7 and M1-M5 belong to the displacement and mixing types, respectively. Detailed locations of the ventilation equipment (the air exhaust, supply and duct) are shown in Fig. 2.

| Configuration | Connecting duct | Description |
|---------------|-----------------|-------------|
| D1            | AR = 50 (1 × 0.02 m²) | Air exhaust on the ceiling is located above a patient. |
| D2            | Air exhaust on the ceiling is located behind a patient. |
| D3            | Smaller exhaust (half of the others); the location is the same as D1. |
| D4            | Chair with 4 legs with the air exhaust at the same location as D1. |
| D5            | 0.7 × 0.0286 m² | Smaller AR (25) of connecting duct; the location of the air exhaust is the same as D1. |
| D6            | 0.4 × 0.05 m³ | The smallest AR (8) of connecting duct; the location of the air exhaust is the same as D1. |
| D7            | Two ducts | Two ducts besides the patient; the location of the air exhaust is the same as D1. |
| M1            | 0.05 × 0.4 m³ | Ventilations near the floor (air exhaust behind a patient) covers the exhaust. |
| M2            | 0.8 × 0.025 m² | Ventilations near the ceiling (the same location of air exhaust as D2) |
| M3            | 0.2 × 0.1 m² | Ventilations near the floor; flow from the duct moves far from the exhaust (the same location of air exhaust as M1). |
| M4            | 0.2 × 0.1 m² | Ventilations near the floor (the same location of air exhaust as M1). |
| M5            | 0.2 × 0.1 m² | Ventilations near the ceiling (the same location of air exhaust as D2) |

Like the hospitals and health care centers but also to be carried inside a vehicle. Considering its usage, it is necessary to separate the doctor’s (positive-pressure) and patient’s (negative-pressure) room, and each should be equipped with a ventilation system (WHO, 2020). In reality, several accessories like the rubber gloves would be installed between the rooms to assist the doctor’s diagnosis; however, they are omitted to draw general flow features in this study. In addition, the doctor manikin was also ignored considering the patient as a source of contaminants and the aerosol removal performance in the negative pressure chamber is our interest.

The model exhibits the dimensions of the actual clinic environment (Fig. 1). The clinic facility includes a curved ceiling with a radius of curvature of 0.35 m. Both the negative and positive pressure rooms have the dimension of 1.48 m × 1.48 m × 1.8 m in the spanwise (x), streamwise (y), and normal (z) directions, respectively. Each unit is connected by a rectangular connecting duct, of which the cross-sectional area is 0.02 m² in all configurations. The cross-sectional area of the apparatus to maintain the negative and positive pressure difference at the exhaust and supply, respectively, is 0.25 m × 0.5 m.

To deal with the questions we raised, we performed computational fluid dynamics simulations, considering various ventilation strategies represented by twelve combinations of air exhaust and connecting ducts (Table 1 and Fig. 2). Considered configurations are further divided into displacement (D series) and mixing (M series) ventilation. So far, no quantitative evaluation of ventilation configuration has been provided based on the multiphase flow investigation regarding the pathogen removal, i.e., flow-particle interaction. It is also necessary to analyze the coupling effect on the ventilation flow caused by the presence and movement of indoor occupants or objects. For the present configuration (Fig. 1), the airflow from the connecting duct first encounters the personnel (patient) seated on a chair and then escapes throughout the exhaust. Therefore, we vary the shape (i.e., aspect ratio) and position of the duct in each configuration, and the distance to the air exhaust (Fig. 2); in all cases, the cross-sectional area of the connecting duct is maintained to be the same. In addition, we examine the (i) effect of different geometry of the furniture; e.g., the design of a chair such as the four-legged and box-shaped ones, and (ii) the change in aerosol exhaust performance according to the air exhaust size. Among two typical types of the ventilation, we focus on the displacement type in more detail, but various mixing types are also analyzed together based on the relationship between duct flow and air exhaust. For the mixing type, we aim to characterize how much the duct flow covers the exhaust. The air supply and exhaust installation location are also varied such as the ventilation system near the floor and ceiling.

We planned a numerical simulation scenario identical to the actual COVID-19 diagnostic test to evaluate ventilation performance in a realistic situation (Fig. 3). As shown in Fig. 3(a), we have divided the protocol into three stages: in stage 1 (patient check-in), the patient enters the negative-pressure unit and undergoes a diagnostic test for five minutes (with the face mask off) and the patient leaves the room in stages 2–3 (patient check-out). In stage 2 (for three seconds), the door opens and closes and then a heat source (patient) is eliminated from the test room in stage 3. During all stages, the ventilation is continuously operating. As a representative example (configuration D7), in stage 1, it is well-simulated that the exhaled aerosols rise toward the ceiling under the influence of buoyant flow from manikin and respiratory flow (Fig. 3b). Here, the aerosols are continuously injected into the testing unit. Since the aerosol injection is halted in stage 2–3, the remaining aerosols migrate under the influence of the door movement and ventilated flow (Fig. 3c). Across the stages 2 and 3, the amount of aerosols remaining in the unit decreases continuously with time. In this context, the simulation timelines were defined according to the presence of aerosol injection; Tini for the stage 1 and Tout for the stages 2–3 (Fig. 3a). As a quantitative indicator to assess the ventilation performance, we defined the pathogen-laden aerosol removal time (τr) as the time taken in the stage 2–3 to reduce the aerosol amount below the threshold (1% of the initial aerosol mass) after the stage 1. Typically, the ACH (air change per hour; the rate at which indoor air is exchanged for the fresh outdoor air) has been adopted as a metric to evaluate the ventilation performance in a confined indoor environment. Thus, it is understood that the ACH is the inverse of the present timescale (τr) (Mathai et al., 2021).

As shown in Fig. 3(c), the entrance door (0.7 (width) m × 1.3 (height) m), opened and closed around a hinge (x, y) = (0.35 m, 0 m), is installed in the center of the backside of the patient in the negative-pressure unit. Inside this three-dimensional model, we located a chair and patient (aerosol-generating source) in the stage 1 and chair-only in subsequent stages. The other interior part of the facility is simplified and ancillary accessories such as the medical gloves that do not critically affect the ventilation are omitted. The patient was made to sit 0.25 m (the minimal distance) away from the wall in consideration of actual treatment. The height of patient, seated, is 1.35 m and the width of the patient is the same as the chair. The cross-sectional area of the chair’s lower part is 0.4 m × 0.5 m and the total height is 1 m. The computer-aided design model for the whole facility and the interior geometry was created with SOLIDWORKS and Design Modeler, an ANSYS Fluent module.

2.2. Boundary conditions and grid system description

To obtain the exact results compared to the actual situation, we elaborated the adequate boundary conditions for the CFD domain. The airflow temperature, velocity, and turbulence intensity are set as input boundary conditions at the air supply and exhaust. For the present simulations, the air was assumed to be an ideal gas and the ventilation flows at the air supply and exhaust were set to have a room temperature...
(20 °C). The air exhaust and supply were designated as the pressure-outlet and inlet, respectively. In order to maintain the pre-determined negative and positive pressures in each unit, the gauge pressures were specified as – 2.5 Pa and 5 Pa, respectively. The unsteady velocity of the respiratory flow at nostrils was established via the user-defined function (UDF) for the velocity-inlet condition as $3.75 \sin (0.4tT_{in})$ (Gupta et al., 2010). In addition, the Boussinesq approximation is considered to account for the buoyancy-driven flows, owing to some heat sources such as thermal manikin (heat flux of 39 W/m²) and light bulb on the ceiling (10 W/m²) (Assaad et al., 2018; Katramiz et al., 2020). As a source of the pathogen aerosols, the respiratory flow (at the initial temperature of 37 °C) is introduced as a buoyancy-driven flow. To mimic the realistic human nasal breathing as much as possible, we apply the experimentally measured human nose shape (nostril size of 0.8 m² and nose angle of 40°) and time-dependent sinusoidal flow rate profile (peak value of 0.3 L/s and period of 5 s) (Fig. 1) (Melikov and Kaczmarczyk, 2007; Katramiz et al., 2020).

For the grid system required for the present simulation, we produced an unstructured, tetrahedral mesh to be applied to the three-dimensional model using the Ansys Fluent module. The mesh consists of $8.5 \times 10^5$ cells with a mean size of 6 cm, orthogonal quality of 0.77 (the minimum value of 0.163), and aspect ratio of 1.85 (the maximum value of 11). The grids near the nostrils are refined to 0.05 cm in order to track the particle trajectory more accurately. Specific grid size was also assigned to characteristic regions with complex geometry or high flow rates; 2.5 cm for the negative pressure, positive pressure equipment and the duct, and 1.5 cm for human body and chair. The grid independence study was conducted in terms of the air mass flux (kg/s) parameter through the air exhaust (see Fig. S1 in the Supplementary Materials). As shown, the air mass flux converges (errors under 1%) as the number of grids increases from the resolution selected in this study. Additionally, the velocity profiles at the air exhaust and the temperature profiles above the manikin obtained with different grid numbers are plotted in Fig. S2 in the Supplementary Materials. As shown, both the velocity and

Fig. 2. Ventilation configurations for the infectious disease treatment (diagnostic test) unit considered in the present numerical simulations. Black arrow indicates the connecting duct between the negative and positive pressure room. Blue and red planes denote the air exhaust and supply. Detailed descriptions are provided in Table 1.
temperature profiles converge as the number of cells increases more than $8.5 \times 10^3$. Therefore, the physical quantities simulated based on the adopted mesh resolution in the present work are independent of the number of grids.

### 2.3. Numerical simulation models for airflow and particle dynamics

Numerical simulations were conducted using the commercial CFD software (ANSYS FLUENT 21 R1) to resolve the airflows and particle (aerosol) trajectories. The continuity and momentum equations of the incompressible Navier-Stokes (RANS) model were solved using the transient Reynolds Averaged Navier-Stokes (RANS) incompressible solver with Re-Normalization Group (RNG) k-ε model. According to previous studies, the RANS approach with the RNG k-ε model showed acceptable computational cost and superior performance compared to other turbulence models compared with PIV results in the indoor airflow field (Evola and Popov, 2006; Bartzanas et al., 2007; Kobayashi et al., 2009). Furthermore, the RNG k-ε model with the scalable wall function (for the purpose of resolving the boundary layer) and full buoyancy effects was employed to simulate the generation and dissipation of turbulent energy. Therefore, the present approach has a good performance in modeling indoor turbulent flow fields (Ramponi and Blocken, 2012; Abuhegazy et al., 2020). On the other hand, the Pressure-Implicit with Splitting of Operators (PISO) algorithm is applied for solving the pressure-velocity coupling owing to its suitability for transient flows (Assaad et al., 2018; Katramiz et al., 2020). The momentum, energy, k, ε, and turbulence were discretized using the second-order upwind scheme. The “PRESTO!” scheme was used for the pressure equation, accurately accounting for the pressure gradient near the boundaries (Assaad et al., 2018; Katramiz et al., 2020). With a transient solver and second-order implicit time-stepping scheme, we used a time step as 0.5 s, which is sufficiently smaller than the period of the nasal respiratory flow. In the literature, we found that previous studies on the similar situations used the time step of 0.5–2 s (Perino, 2009; Wang et al., 2017). The reason for using the fully implicit scheme is that it is unconditionally stable for time-step size. The solution was considered convergent when the scaled residuals reach $10^{-4}$ for all quantities except energy, which should be less than $10^{-6}$. The simulation of each case consumed roughly two days of computational time using sixteen CPU cores.

Once the continuous phase solution converges at each time step, the trajectory of the discrete phase (pathogen-laden solid aerosol) is tracked through the Lagrangian technique (DPM) based on the airflow solution. It has been reported that the human exhalation droplet size spans a wide range of 0.01–1000 μm; however, we only consider the aerosol particles $< 5 \mu m$, which will follow the airflow faithfully (WHO, 2014; Bake et al., 2019; Fennelly, 2020; Milton, 2020). It is because the droplet aerosols of $O(10 \mu m)$ evaporate within 0.1–1 s (Redrow et al., 2011) and those of $O (100 \mu m)$ are directly settled on the floor or other nearby surfaces (Liu et al., 2017). For this reason, in reality, the median size of most long-lived exhaled aerosols is between 0.7 and 1.0 μm (Bake et al., 2019) and most of the aerosols (87%) with influenza viral RNA such as COVID-19 are found to be less than 1 μm in size (Fabian et al., 2008; Fennelly, 2020). Therefore, based on the experimentally measured virus-laden droplet nuclei size (0.1–1.5 μm (Fennelly, 2020)), we consider the solid (1000 kg/m$^3$) spherical aerosols with an average diameter of 0.9 μm, of which the size distribution is given by the Rosin-Rammler model. As noted, we consider that the present dispersed phase does not evaporate further. Based on the Rosin-Rammler model, the mass fraction of particles whose diameter is greater than $d$ is defined $Y_d = e^{-d/d_o} / (d_o \sqrt{2 \pi})$, where $d_o$ is the average diameter; $n$, spread parameter. To set approximately 80% of the aerosols to be smaller than 1 μm, $n$ was chosen to be 5 (i.e., sharp distribution profile). The considered size distribution of the aerosol is shown in Fig. S2 in the Supplementary Materials. These solid aerosols were released through the nostrils at a mass flow rate of $3.4 \times 10^{-13}$ kg/s (matching the experimentally measured breathing flux) and dispersed in the test room (Fennelly, 2020). Given the low...
concentration of aerosol (aerosol volume fraction in the test room \( \ll 10^{-6} \)), the effect of the particles on the flow of air is negligible, i.e., one-way coupling between the aerosol and airflow (Abuhegazy et al., 2020). On the other hand, Belosi et al. (2021) showed that the interaction between the emitted virus-laden aerosols and the pre-existing atmospheric particles has a very low probability, which would not be able to change the dynamics of virus-laden particles, under the typical condition (size and concentration) of particles. Thus, in the present simulation, the interaction between the aerosols and the pre-existing particles was ignored.

The equation of motion for the particle is generally given in the form

\[
d\vec{v}_i/dt = \vec{F}_{d} + \vec{F}_{LS} + \vec{F}_g + \vec{F}_a + \vec{F}_{basset} + \vec{F}_{brownian},
\]

where \( \vec{v}_i \) is airflow velocity, \( \vec{F}_d \) is the Stokes drag modification function (\( \vec{F}_d = 1 + 0.156Re_p^{0.87} \)), \( Re_p \): particle Reynolds number, and \( \tau_p \) is the particle response time (\( \tau_p = \rho_p d_p^2 C_d/(18 \mu) \)), \( \rho_p \): solid particle density; \( d_p \): particle diameter; \( \mu \): fluid viscosity). Additionally, the effects of turbulence on the particle dispersion were considered through the discrete random walk method implemented in ANSYS FLUENT. The discrete phase boundary conditions type is established as follows. The trap (adhesion) condition was applied for solid walls (including the curved ceiling), a chair, and manikin so that the aerosol trajectory calculation was terminated when it reaches the nearest grid cell from the wall. For the flat ceiling, the reflection condition was defined considering the airflow from the air exhaust and supply. Finally, the escape condition was applied to the air exhaust, supply, and human nostrils.

\( \theta_0 = 151 \text{ s} \) for the configuration D1 (a) and D4 (d). Here, the aerosol distribution in the whole three-dimensional chamber is reflected on the y-z plane (\( x = 0 \text{ m} \)), and the color denotes the aerosol residence time in the room.

**2.4. Qualitative validation of the turbulence model**

The turbulence model we used for simulating the airflow pattern (including the temperature and velocity fields) was validated by comparing the thermal plume from the body with previous studies for indoor airflow (see Fig. S4 in the Supplementary Materials). If the mass flux of the ventilation flow is large, it will inevitably affect the behavior of the buoyant thermal plume. Therefore, the detailed characteristics of the thermal plumes are slightly different from each other, owing to the different indoor ventilation conditions (e.g., the air change rates). However, it is noted that the present results share the physically important features with the previous CFD and experimental data. For
example, the thermal plume tends to ascend above the manikin head while the warmer breath jet flow is ejected through the nostrils. The velocity magnitude of the ascending plume is 0.15–0.25 m/s in the present study, which is comparable to the plume velocity observed with previous CFD (Salmanzadeh et al., 2012) and experiment (Li et al., 2017). In addition, the temperature of the present thermal plume above the manikin head is 24–26 °C, which also agrees with the previous CFD results (Voelker and Alsaad, 2018; Liu et al., 2019).

3. Results and discussion

3.1. Interaction between human breath and the flow inside the unit

3.1.1. Aerosol behaviors and its removal mechanism

Although the duct connects the positive and negative pressure units, their internal pressure is always maintained to be 5.0 Pa and −2.5 Pa, respectively (Fig. 4a). While holding the differential pressure across the units, the airflow is continuously entrained into the negative pressure chamber from the positive-pressure side through the connecting duct; therefore, there is no inflow of aerosols into the positive pressure unit. Since this duct is installed in front of the patient in all the ventilation configurations (Figs. 1 and 2), the duct flow into the negative pressure unit directly interacts with the patient’s respiratory flow. As illustrated in Fig. 4, even though both cases have the same connecting duct and displacement ventilation type, the dynamics of aerosol vary significantly depending on the indoor objects. This is because the main stream induced by the duct flow, responsible for the interaction with the respiratory flow, is different depending on the geometry of the chair on which the patient sits. With a box-type chair (D1), for example, the duct flow encounters the respiratory flow above the chair and then sucked up straight to the air exhaust (Fig. 4a). Accordingly, the aerosol cluster contained in the respiratory flow is discharged directly above the patient without being scattered (or dispersed) widely (Fig. 4b). While the major flow structure is set from the duct to the air exhaust, pushing the aerosol cluster upward, the duct flow also goes around the chair forming a pair of recirculating bubbles (regions) behind the patient (Fig. 5a). This vortical structure sweeps the entrained aerosols from center of the vortex towards the air exhaust at a relatively higher velocity (i.e., flows toward the patient; see blue arrow in Fig. 5a), which prevents the aerosol dispersion. Thus, it is understood that all the flow structures evolving from the duct flow push the respiratory flow toward the air exhaust in all directions throughout the stage 1, reducing the flow mixing and increasing the aerosol discharge rate. Conversely, with the chair having four legs (D4), most of the airflow from the duct passes under the chair (i.e., flows away from the patient; see blue arrow in Fig. 5b) and along the wall behind the patient (Figs. 4c and 5b). As a
result, the respiratory flow is entrained into the flow from the duct and some aerosols rise upward along the wall together (Fig. 4d). Since there is no organized flow structure to suppress the aerosol dispersion, most of the aerosols descend and disperse throughout the negative pressure unit. Consequently, the aerosols are concentrated on the bottom of the unit and less amount rises to the air exhaust. As shown at \( T_{in} = 150 \) s, the aerosol mixing in the room is greatly enhanced over time in the case of D4, but it is preferentially concentrated around the air exhaust in the configuration D1 (Fig. 4b and d).

Unlike the displacement type ventilation, there is no immediate interaction between the internal flow and human breath for the mixing type. Fig. 6 shows the cases where the airflow from the duct partially or does not cover the air exhaust among the mixing-type ventilation (M3-M5). The respiratory flow is entrained into the airflow from the duct and directed straight to the opposite wall, not toward the exhaust, with a velocity as high as 2.0–3.0 m/s, and the aerosols are also concentrated around that flow structure (Fig. 6a). Although the airflow passing around the exhaust acts as an ‘air curtain’ to block the transport of the particles to the exit, it can somehow contribute to the enhancement of the ventilation since the fast airflow around the corner of the chamber can transport the aerosols quickly to the vicinity of the exhaust. Furthermore, the buoyancy effect by the respiratory flow improves the ability to remove aerosols in the air. Even though the aerosols are in general dispersed more uniformly inside the compartment under the mixing-type ventilation (Fig. 6b), the aerosols tend to rise toward the ceiling by the buoyant respiratory flow (warm plume) and merge with the airflow from the duct to be guided to the exhaust located near the ceiling. Therefore, most of the aerosols are preferentially collected in the upper part of the negative-pressure unit. If the ventilation system (duct and exhaust) is located near the floor (Fig. 6c), the aerosols are divided into a descending group toward the flow from the duct and ascending group so that they begin to mix evenly inside the unit. As the time passes further (\( T_{in} = 100 \) s, for example), the flow mixing is enhanced and the compartment is filled with a high concentration of aerosol (Fig. 6b and c).
3.1.2. Quantitative analysis of the pathogen-laden aerosol removal

Based on the qualitative analysis for the airflow structure and corresponding kinematics of the aerosols in the above, we evaluate the virus removal performance using a quantitative index and understand the mechanism of aerosol behavior. Fig. 7 maps the capability of the aerosol particle removal of tested ventilation configurations with (stage 1) and without (stage 2–3) the patient (i.e., source of the aerosol and heat flux). For each ventilation configuration, we calculated the amount of aerosols (kg) remaining in the negative-pressure unit after the stage 1 than D1. Nevertheless, they are also classified two regimes in the stage 2–3 depending on whether the particle dispersion is inhibited further or the flow mixing is enhanced during the ventilation. In the regime 1 (such as the configuration of D1), the major flow structure inside the unit is the divided flow from the duct, which climbs up along the walls and reaches near the air exhaust, while pushing and entraining the aerosols (Fig. 4a). Here, the most distinguished flow structure in the regime 1 is the organized airflow structure surrounding the respiratory flow, with which the airflow suppresses the respiratory flow diffusion and causes the aerosol cluster to agglomerate under the exhaust. Since the cases of D2 and D3 have less favorable conditions for the particle removal such as the air exhaust located behind the patient and smaller exhaust size, respectively, they show quite lower ventilation efficacy (larger number of remaining particles) in the stage 1 than D1. Nevertheless, they are also equipped with a slit-duct (like the case of D1), through which the airflow forces the aerosols to be rapidly discharged to the outside following the mechanism of regime 1, and thus they achieve higher ACH values (in the stage 2–3) than other cases. Similarly, the mechanism of the regime 1 is not fully achieved by the condition of D5 owing to the lower duct AR (= 25) than D1 (AR = 50), so that the ventilation performance was relatively lower during the aerosol-generating process (stage 1) (more detailed analysis will be given in the next chapter).

On the other hand, for the regime 2, even it belongs to the displacement type, the airflow from the duct barely controls the aerosol clustering along all of the walls of the compartment, pushing up the aerosol clusters simultaneously (see Fig. 4a). Large wake vortex pairs behind the chair also contribute by transporting the dispersed aerosols in the unit to the air exhaust. Due to this particle-flow interaction, the aerosol cluster near the exhaust is discharged quickly, i.e., having the shorter residence time (Fig. 7). After the patient leaves, the major flow structure inside the unit is retained; the airflow continuously climbs up along the walls driving the particles upward, and thus the aerosols rise up at a high velocity (> 0.4 m/s) and are rapidly discharged outside (Fig. 8a). On the other hand, for the mixing type, the airflow from the duct tend to only sweep the particles locally so that the aerosol discharge rate is relatively lower while the indoor aerosol dispersion is strengthened.

In Fig. 7, it is also noted that some of the displacement-type ventilation have a lower ACH than the mixing-type cases in the stage 2–3. In the above, we have found that the high-performance displacement ventilation systems are designed so that the duct flow is divided along several directions, which rise up along the side walls of the chamber. As a result, the airflow encircles the buoyant respiratory flow and pushes the aerosol cluster towards the air exhaust. In contrast, the other systems only allow the airflow from the duct to move locally within the unit. This type of duct flow is not capable of controlling the respiratory flow and thus cannot minimize the dispersion of aerosol cluster. In this context, we additionally classified two regimes in the stage 2–3 depending on whether the particle dispersion is inhibited further or the flow mixing is enhanced during the ventilation. In the regime 1 (such as the configuration of D1), the major flow structure inside the unit is the divided flow from the duct, which climbs up along the walls and reaches near the air exhaust, while pushing and entraining the aerosols (Fig. 4a). Here, the most distinguished flow structure in the regime 1 is the organized airflow structure surrounding the respiratory flow, with which the airflow suppresses the respiratory flow diffusion and causes the aerosol cluster to agglomerate under the exhaust. Since the cases of D2 and D3 have less favorable conditions for the particle removal such as the air exhaust located behind the patient and smaller exhaust size, respectively, they show quite lower ventilation efficacy (larger number of remaining particles) in the stage 1 than D1. Nevertheless, they are also equipped with a slit-duct (like the case of D1), through which the airflow forces the aerosols to be rapidly discharged to the outside following the mechanism of regime 1, and thus they achieve higher ACH values (in the stage 2–3) than other cases. Similarly, the mechanism of the regime 1 is not fully achieved by the condition of D5 owing to the lower duct AR (= 25) than D1 (AR = 50), so that the ventilation performance was relatively lower during the aerosol-generating process (stage 1) (more detailed analysis will be given in the next chapter).

On the other hand, for the regime 2, even it belongs to the displacement type, the airflow from the duct barely controls the aerosol...
cluster dispersion but rather enhances mixing inside the unit. Thus, among the dispersed aerosols, only those nearby the airflow are entrained and discharged. According to this flow structure, the ventilation systems in the regime 2 yield lower ACH values than those in the regime 1. The airflow for the case of D4 (Fig. 4b) and D7 reaches the exhaust by sweeping the dispersed aerosols near the wall, which does not help the upward movement of the respiratory flow (including aerosol cluster). In this procedure, indoor aerosol dispersion is intensified over the entire area. Additionally, for the case of D6 having the smallest duct AR (= 8), the airflow from the duct is not divided in front of the chair but rises immediately and exits through the exhaust (see Fig. 13a). Consequently, only the aerosols entrained in the airflow in the front of the duct move upward and are discharged, but those in other areas are stationary (Fig. 8b).

Based on the criterion to classify the regimes, it is obvious that the mixing-type ventilations would belong to the regime 2 (Fig. 7). Among the mixing types, on the other hand, the configurations of M5 and M4, in which the airflow from the duct partially covers the exhaust opening, are capable of removing the aerosols relatively quickly. Thus, even if the airflow may act as an ‘air curtain’, the ventilation performance can be enhanced by designing the duct narrower than the exhaust. Otherwise (if the airflow from the duct is directed far away from the exhaust, such as the case of M3), most of the aerosols are just dispersed inside the compartment, decreasing the aerosol discharge rate (Fig. 6a). Although the ‘air curtain’-like flow structure may help discharging the aerosol, the duct flow that completely covers the exhaust opening is quite disadvantageous for aerosol discharge (M1 and M2). Although the flow structure of partial ‘air curtain’ can improve the mixing ventilation performance corresponding to regime 2, it is shown that the aerosol discharging performance of regime 1 is much better.

In short, the airflow of the regime 1 suppresses the aerosol dispersion but it rather accelerates indoor aerosol mixing in the regime 2. To support this analysis, we additionally compared the mass fraction of escaped (through the air exhaust) and trapped (deposited) aerosol to the injected aerosol mass in the negative-pressure unit for different ventilation configurations in the stage 1 (Fig. 9). Here, the total trapped particle mass was calculated as the sum of the mass of particles whose trajectory ends on the wall. Since the airflow of D1 (regime 1) transports the aerosol cluster included in the respiratory flow to the air exhaust with less dispersion, twice as much aerosols are discharged to the exhaust than other configurations (M4, D4 and M1 belonging to the}

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**Fig. 9.** Comparison of the escaped aerosol mass fraction and trapped aerosol mass fraction to the injected aerosol mass in the negative-pressure unit in stage 1 (D1, M4, D4 and M1; from highest to lowest ACH).

**Fig. 10.** Aerosol distributions and airflow velocity vectors at the stage 1 (only the negative-pressure unit is shown): (a, b) configuration of D2 (exhaust is located behind a patient); (c, d) configuration of D3 (smaller exhaust). For each configuration, the aerosol distributions (three-dimensional data are reflected on the y-z plane (x = 0 m)) are shown for $T_{m} = 12$ s and 156 s; the color denotes the aerosol residence time in the room. The velocity fields are shown on the y-z plane (x = 0 m).
Therefore, the amount of aerosol deposited on the wall is quite small. However, in the regime 2, the portion of trapped aerosol mass increases over that of discharged one. This indicates that the airflow plays the role of scattering the aerosol widely in the room, rather than dragging it to the exhaust, which agrees with our understanding such that the more intense the aerosol mixing, the lower ACH (ventilation performance).

3.2. Critical factors to improve the aerosol exhaust

So far, we have analyzed the pathogen-laden aerosol removal performance of various configurations qualitatively and quantitatively, and based on this, we suggested that the effective suppression of indoor aerosol mixing is the key mechanism to achieve the complete and fast aerosol removal. Followingly, in this chapter, we discuss the critical factors that control the indoor aerosol mixing.

3.2.1. Air exhaust location for displacement types

In Fig. 10, two configurations (D2 and D3) with a long-slit duct are compared in terms of aerosol distribution and airflow velocity fields in the stage 1. As we have explained, for both cases, the airflow from the duct is strong enough to proceed behind the patient after being divided around the chair. Similar to the ventilation mechanism of D1, the divided flow climbs up the walls of the negative-pressure unit and merge under the air exhaust (Fig. 10b and d), and the aerosol cluster is formed in the upper part of the room (Fig. 10a and c). However, for the case of D2, the entrained aerosol cluster should travel a relatively long distance to the air exhaust, and some are dispersed inside the compartment during the migration. In addition, the entire cluster cannot be forced to move just below the exhaust, so it is distributed over a broader range, compared to the cluster distribution of the case of D1 (compare Figs. 4b and 10a). For the case of D3, the area of the exhaust is reduced by 50% from other cases, and its disadvantageous influence on the particle discharge is manifest in Fig. 10b, despite it operates with the same mechanism as D1. Since the air exhaust for D1 case is wide enough to accommodate the aerosol cluster, the cluster is less dispersed and discharged quickly through the exhaust (Fig. 11a). With the reduced exhaust area (D3), the respiratory flow from the patient impinges on the ceiling scattering the aerosols behind the patient, and only a fraction of the aerosol cluster is eventually discharged (Fig. 11b). Therefore, the cases of D2 and D3 show relatively low particle removal rates in the stage 1, owing to the differences (from the optimal case of D1) in the air exhaust design. The present results make it clear that the higher ACH can be achieved through the strong and coherent airflow from a slit duct, and we further demonstrate that not only the existence of the airflow toward the exit itself is essential, but...
it must also be formed to cover the entire compartment (e.g., flow along the walls) to prevent the particle mixing or localized concentration (trap).

### 3.2.2. The optimized condition for the airflow from the duct

To examine the effects of aspect ratio and location of the duct on the suppression of aerosol dispersion, we have compared different AR’s and arrangement in Fig. 12. When AR is reduced to 25 (D5), the ACH value is slightly dropped from that of D1 (AR = 50) because the mass flux of the airflow behind the patient decreases (Fig. 12a and b). Although the aerosol cluster is formed in front of the patient, the less coherent airflow structure results in the unstable, i.e., asymmetrically scattered, aerosol cluster (Fig. 12b). As shown, the particle dispersion is rather enhanced owing to the relatively weaker dragging force behind the patient. In this context, it is preferred to design the ventilation system to achieve the airflow passing behind the patient (case D7 in Fig. 12c) rather than just rising in front of the patient (case D6 in Fig. 12d). It is clear that both the divided flows in front and rear of the patient are essential, but the particle dispersion motion is encouraged without the flow pushing the aerosol cluster behind the patient. For the case of D6, the airflow from the duct immediately pushes the aerosol cluster toward the exhaust, but at the same time, the aerosols are dispersed around the patient (Fig. 12d). This is because there is no airflow or coherent vortical structure behind the patient which pushes the respiratory flow and carries the aerosols to the exhaust. Since there is no organized airflow structure to transport these dispersed aerosols up to the air exhaust (Fig. 13a), there are stagnant aerosols everywhere inside the compartment except just in front of the duct (Fig. 8b). With the ventilation system that discharges only the aerosols entrained into the airflow located in the localized area, the aerosols are removed very slowly, resulting in the lowest ACH value among the displacement types (Fig. 7). On the other hand, the flow structure in the case of D7 does not directly encounter the respiratory flow, but results in a relatively higher ACH value (Fig. 12c). For this case, two rectangular ducts are located at both sides of the patient and thus the airflow from the duct only goes around the chair not in front of the patient (Fig. 13b). The duct flow eventually produces the vortex pairs behind the patient (also noted in Fig. 5a) and also climbs up along the walls. Since the duct flow continuously pushes the aerosol cluster behind the patient, the cluster does not collapse quickly despite the entrainment of the aerosol into the airflow. During the aerosol-generating process, the aerosols are not discharged immediately so that the concentration of the cluster increases (higher remaining particle number in the stage 1; see Fig. 7). When the aerosol generation ceases (stages 2–3), however, the aerosols in the cluster continuously rise to the air exhaust and are discharged. In particular, the aerosol removal rate increases because the dispersed aerosols far from the air exhaust (i.e., having low chance to be discharged) are rapidly transported by the airflow (Fig. 8c). Since the entire duct flow evolves into the large vortices and rising airflows behind the patient, it can entrain more aerosols than in other cases and transport them faster even if the duct flow is localized only behind the patient.

In summary, as illustrated in Fig. 12, the presence of divided duct flow into two directions (in front of and behind the patient) inside the entire compartment contributes to the improvement of ventilation mostly. Based on this comparison, if there is a practical limitation (such as the position of equipment and furniture) preventing a slit duct at the bottom, it is recommended to have ducts on both sides of the patient (Fig. 12c). This ventilation system will allow the airflow behind the patient to minimize indoor aerosol cluster collapse and to increase the aerosol removal rate. In this context, inducing the airflow (which moves from bottom to top) surrounding the respiratory flow so as not to disturb the aerosol cluster in a displacement ventilation system is the most significant ventilation mechanism that can be utilized in general.

### 3.2.3. Effect of door motion (opening and closing)

In addition to the issue of ventilation system design, we found that the opening and closing of the door after the aerosol-generating process significantly affects the aerosol removal time. In the present configuration, the door of the negative-pressure unit is designed to move from the closed ($\theta = 0$) to fully opened ($\theta = \pi/2$) state for 1.5 s (Fig. 14) and returns to the closed state for next 1.5 s (Chang et al., 2016). The angular velocity of the door movement is modeled as $(\pi/3)\cos(\pi/3)T_{out}$ and

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**Fig. 12.** Aerosol distributions on the $y$-$z$ (top) and $x$-$y$ (bottom) planes (only the negative-pressure unit is shown) at the end of the stage 1 ($T_{in} = 300$ s): (a) D1; (b) D5; (c) D7; (d) D6. For each configuration, the aerosol distributions of the three-dimensional data are reflected on the $y$-$z$ plane ($x = 0$ m) and $x$-$y$ plane ($z = 0$ m), respectively; the color denotes the aerosol residence time in the room. Black arrows denote the directions of the airflow from the duct and exhaust, and the orange arrows denote the airflow directions and their length denotes flow velocity (not drawn to scale).
\[ \pi/3 \cos(\pi/3) T_{\text{out}} \] for the opening and closing, respectively (Hang et al., 2015). During the stage 2, the amount of aerosol in the negative-pressure chamber decreases by 10–40%, compared to the amount remaining immediately after the stage 1 (Fig. 15). It is noted that the aerosol emission efficiency by the door motion is proportional to the ventilation performance in the stage 1. In other words, the ventilation system with a superior aerosol discharging performance in the stage 1 is assisted more by the door sweeping motion. Since the air mass flux at the exhaust is the same for all configurations (see the inset of Fig. 15), door motion does not reverse the trends that we have shown in Fig. 7. Within a very short duration (three seconds) of the stage 2, the aerosol behaviors did not vary much and maintained the same as in the stage 1, while the discharge rate increased sharply. This is because the air mass flux discharged from the exhaust increases drastically by about three times. During the initial door opening \( T_{\text{out}} = 0–0.5 \text{s} \), indoor air goes out through the door so that the outdoor air flows into the room through the exhaust (Fig. 14 and S5 in the Supplementary Materials). Later, substantial airflow is introduced into the room through the open door until it is closed, resulting in much greater discharge (compared to the stages 1 and 3) through the exhaust. Thus, it is understood that even a short door motion can be also strategically useful in removing the considerable amount of pathogen-laden aerosols through the exhaust.

**4. Conclusion**

In the present study, we investigated the dynamic movement (dispersion and discharge) of the aerosols ejected from the patient (representing the hazardous pathogen-laden breathing flow), governed by the particle-flow interaction with the airflow field from the ventilation system. To characterize the optimal mechanism for fast and complete removal of aerosols, we performed rigorous CFD simulations for the multiphase flows inside a negative-pressure clinic compartment with various ventilation configurations. To simulate the procedures of utilizing the facility as the actual diagnostic test (such as the COVID-19 testing), we planned the scenario as three stages; first, the patient stays in the test room for five minutes in the stage 1 and in the stage 2, the patient exits the unit with a door movement. Finally in the stage 3, the ventilation continues without the patient. That is, the aerosol-generating procedure only takes place in the stage 1 and only the ventilation is activated in the stages 2–3 with no more aerosol production. In this context, this study aimed to find the optimized mechanism to achieve the highest pathogen-laden aerosol emission rate in all stages, and consequent system design.

Being distinguished from the existing condition (BSL-2) suggested by the WHO (WHO, 2020), the present ventilation strategy is innovative such that the negative-pressure unit where the diagnostic test is performed is connected to other spaces. While the typical diagnostic
workplace in a pandemic situation is completely isolated from the outside space, the open duct connects the present negative-pressure unit to the positive-pressure unit. Therefore, the airflow in the positive pressure side is entrained into the negative-pressure side through the duct, and the subsequent airflow structure determines (improve or degrade) the performance of the ventilation system. Assisted by the connecting duct, all ventilation configurations in this study showed the aerosol removal performance at least twice better than the conventional closed chambers (CDC, 2003; Weiland et al., 2021). Especially, the case of D1 (optimized condition) removed 99% of aerosols in the chamber within 4.5 min, which is 4–8 times faster than the previous results from single chamber design. As we analyzed the detailed flow structure inside the negative-pressure unit, we think that the enhanced performance is attributed to the well-organized flow path, carrying the pathogen-laden aerosols, from the connecting duct to the exhaust. Thereby, it is also possible to minimize the unwanted dispersion of the aerosols. This kind of flow optimization is not easily achieved in the single closed chamber. In addition, the continuous airflow through the duct ensures that no aerosols escape (leak) into the positive-pressure chamber.

We demonstrated that it is the key to control the airflow structure inside the unit to improve ventilation performance, by which the aerosol does not disperse immediately after it is sprayed from the patient and stays in the form of a cluster. In this study, the displacement-type ventilation generally exhibits a better ventilation performance than the mixing types, similar to the results of previous studies. On the other hand, there were cases (such as D4, D6, and D7) in which the ACHs are similar to or lower than the mixing type, among the displacement types. We found that this because the ventilation system design influences the flow structure inside the unit and the particle discharge rate varies accordingly; i.e., the aerosol cluster is dispersed or more agglomerated. Therefore, to complement the existing criteria to define the ventilation performance, we additionally defined two regimes. In the regime 1, the airflow from the duct encounters a (box type) chair on which the patient sits interacts with the respiratory flow from the patient and move upward along the walls of the compartment. At the same time, the flow around the chair and patient surrounds the aerosol cluster restraining the indoor aerosol dispersion. Driven by this flow structures, the aerosols are discharged very quickly as the flow pushes the cluster directly up toward the exhaust. On the other hand, in the regime 2, the airflow

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**Fig. 14.** The airflow velocity vector fields for configuration D1 during stage 2 (only the negative-pressure unit and outdoor space are shown). The time-dependent vector fields are shown on the x-y plane (z = 0.3 m). The vector color denotes the velocity magnitude.

**Fig. 15.** The ventilation performance during the stage 1 (with patient) and 2 (door moving without patient) depending on the ventilation configurations. The performance during the stage 2 was calculated as the ratio (%) of aerosol mass removed to the residual aerosol mass after the stage 1. The inset shows the air mass flux through the air exhaust during the stage 2.
can not retrain the respiratory flow accompanying the aerosol cluster, and accordingly, the aerosols can be discharged only effectively when directly entrained to the airflow nearby. That is, the aerosol mixing is accelerated by the airflow in the regime 2, by which the dispersed aerosols in the area where the airflow does not affect are hard to be discharged. Thus, the ventilation performance in the regime 2 is significantly lower than that of the regime 1. To further improve ventilation performance, we found that it is effective to install a duct with a larger aspect ratio at the bottom of the negative-pressure unit. As it becomes large, the airflow with a sufficient mass flux is induced to prevent the dispersion of the aerosol cluster.

For the general indoor community environments (such as a coach bus, high-speed train and health-care center), it is required to consider the ventilation configuration optimized for specific indoor geometry and removal of air contaminants. For instance, the downward ventilation is commonly equipped on the high-speed train (Yang et al., 2018). However, this airflow pattern collapses the aerosol cluster within the breathing flow from the infectious person and tends to disperse the aerosols away from the source. Rather, the downward ventilation is adequate to the intensive care unit and cleanroom facility because it is focused on supplying the fresh cold air to patient’s body (wound). For a general coach bus, the collapse of aerosol cluster (from the infected patient) and spread were observed, as well (Yang et al., 2020).

Compared to these airflow patterns, the airflow from the optimized ventilation configuration in the present study strictly prevents the aerosol dispersion from the breathing flow and rapidly discharges the aerosol through the air exhaust.

While the present numerical simulation model and considered scenarios were designed to faithfully mimic the real environments dealing with the pandemic situation like the COVID-19, there are some issues to be resolved in the future. For example, in this simulation, only the solid super-spread of the virus. In this study, we suggest the ventilation would be different depending on the various breathing maneuvers (Bake with the pandemic situation like the COVID-19, there are some issues to be resolved in the future). For example, in this simulation, only the solid super-spread of the virus. In this study, we suggest the ventilation would be different depending on the various breathing maneuvers (Bake et al., 2019), it would be also necessary to analyze the airflow structure and corresponding aerosol interaction depending on the breathing patterns like the speech and cough. Nevertheless, the ventilation system configuration presented in this study can be applied to various hospital (indoor and outdoor) environments dealing with the airborne infectious diseases. Finally, we hope that the present ventilation strategy can play a major role as a core procedure in the rapid first aid and treatment protocols.

**Statement of environmental implication**

Highly contagious respiratory disease pathogens (including SARS-CoV-2) are significantly hazardous airborne substances because they can be transmitted to the other hosts without physical contacts and they survive for three hours in the unventilated indoor environment. Specifically, hospital environments such as negative-pressure disease diagnostic test or clinic rooms with poor ventilation are considered to the super-spread of the virus. In this study, we suggest the ventilation configuration (including inlet/outlet duct location and shape) with the fastest removal performance of such pathogen-laden aerosols. We believe that the ventilation strategy can play a major role to prevent massive transmission in the community.

**CRediT authorship contribution statement**

Jooyeon Park: Conceptualization, Methodology, Investigation, Writing – original draft. Kwang Suk Lee: Conceptualization, Supervision. Hyungmin Park: Conceptualization, Supervision, Writing – review & editing.

**Declaration of Competing Interest**

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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**Appendix A. Supporting information**

Supplementary data associated with this article can be found in the online version at doi:10.1016/j.jhazmat.2022.128978.

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