

ASSESSING INFECTIOUS DISEASE TRANSMISSION RISK FROM A FLUID MECHANICS PERSPECTIVE: QUANTIFYING THE ROLE OF TEMPERATURE AND RELATIVE HUMIDITY ON EXHALED DROPLET TRANSPORT AND EVAPORATION

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ABSTRACT

Infectious diseases are primarily transmitted through droplets exhaled during respiratory events. In the present work, the risk of infectious disease transmission is evaluated from a statistical analysis based on respiratory droplet trajectory spatial distribution. Trajectories are collected from a fluid dynamics analytical model solving the governing equations of droplet transport, evaporation, energy, and species recently developed and validated by the authors. The analytical model is completed by a droplet release randomisation model, an unsteady empirical model of the respiratory cloud motion within which the droplets are released, and a discrete random walk approach to account for the droplet turbulent dispersion. Imposing the proper boundary and initial conditions, different respiratory events may be modelled with ease and a large number of trajectories can be tracked with a limited computational effort. With assumptions on the exhaled droplet size distribution and viral content, the data collected may be post-processed to provide spatial distribution transmission risk maps able to discriminate between the various transmission routes. In this work the coughing scenario is considered. A number of simulation runs, made of more than twenty thousand droplets each, are performed and discussed to assess the role of ambient temperature and relative humidity on the droplet evaporation, trajectory, and inhalation transmission risk. It is found that airborne risk is largely affected by the ambient thermo-hygrometric conditions while direct inhalation and fomite risks are not. Airborne infection risk and droplet critical diameter are found to almost double passing from cold humid environments to hot and dry ones.

KEYWORDS: Infectious disease transmission, Respiratory droplets, Droplet transport, Virus concentration, Risk assessment, Thermo-hygrometric conditions

1. INTRODUCTION

Pathogen-laden respiratory droplets are among the main means of infectious disease transmission. Different respiratory droplet-based transmission routes are possible, including direct and indirect (airborne) inhalation, and direct and indirect (fomite) contact transmission [1]. Direct inhalation occurs when infectious exhaled droplets are directly inhaled by a susceptible individual due to the small interpersonal distance kept, airborne or indirect inhalation route instead is due to small droplet nuclei that remain suspended in the air for long time and can travel large distances. Contact transmission is referred to as direct when it involves the direct contact between individuals after contamination from droplets, *e.g.* hand-shaking, and indirect or fomite when it occurs through an infected surface, *e.g.* a door-knob.

The size of the droplets exhaled during respiratory events spans over different orders of magnitude, going from ≈ 1 mm down to sub-micron scale [2]. The fluid dynamic behaviour of droplets of such different size varies in terms of transport and evaporation, since different is the balance between gravity and drag forces, and different

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is the surface exposed to convection mass transfer with respect to the droplet volume. Large droplets are mostly influenced by inertia and fall to the ground before evaporation is complete, thus boosting fomite transmission. Small droplets, instead, are driven by drag and evaporate rapidly leaving airborne droplet nuclei made of salts, enzymes, and eventually viruses, that are at the basis of airborne transmission. A critical droplet diameter of $\approx 90 \mu\text{m}$, discriminating between these two behaviours, is generally reported in the literature [3]. This threshold may change moderately depending on the type of respiratory event investigated but is largely affected by the thermo-hygrometric environmental conditions impacting on the droplet evaporation history.

The experimental characterisation of the exhaled droplets transport for different respiratory events is extremely challenging. Some characterisation of the breath cloud has been presented in the literature [4, 5]. While this is important for modelling small droplet transport in the early stages after exhalation, a significant variability in the data has been found between individuals. Much more difficult is the characterisation of the exhaled droplets in terms of size distribution due to their small size, the large range of sizes at stake that cannot be caught by a single instrument, the very rapid evaporation of small droplets after exhalation, and, once again, the variability among individuals. Some attempts have been presented in the literature [6, 7], but the data collected is largely fragmented and the agreement between different works limited [8]. Last, an experimental assessment of the droplet evaporation rates along their trajectory is essentially impossible.

Due to these difficulties, most works tackle the problem using a numerical approach. These can be subdivided into two categories: the works recurring to Eulerian-Lagrangian Computational Fluid Dynamics (CFD) simulations, and those where the analytical equations governing the transport and evaporation of the droplets are mixed with empirical models and solved numerically. The formers are very versatile in their ability to model complex physical systems and are used mostly to present case specific indoor infection transmission scenarios, *e.g.* in an office [9], a supermarket [10], or a car [11]. The high computational cost of the Lagrangian particle tracking, nonetheless, seriously limits the number of particles that can be addressed. The latters are much less computationally intensive and allow a large number of droplets to be tracked with limited effort. Further, being based on a meshless approach they do not depend on the quality of a grid spatial resolution, and the analytical approach grants a high accuracy of the numerical solution. On the contrary, the price to pay is a lesser versatility in modelling complex physical systems, the inability to model particle interactions that yet have been demonstrated to be negligible in respiratory events [12], and the need for a series of empirical models to account for physical features needed to provide the time-varying boundary conditions encountered by the droplet during its flight within the breath cloud. Different empirical models of breath cloud have been proposed in the literature ranging from steady jets [13], to jets having two different spatial development phases (namely, jet and puff) [14], to fully unsteady models [15]. Some model include the effect of buoyancy [16], or implement the effect of turbulent dispersion following a discrete random walk approach [17]. All these models, although elaborated and reasonably accurate, suffer from the fact that the fluid dynamics of buoyant, intermittent, and unsteady exhaled jets is inherently three-dimensional, and cannot be caught correctly by empirical models.

Recently, the authors have developed an analytical model of exhaled droplet transport and evaporation embedding all the most advanced features proposed in the literature [18]. The model addresses the droplet energy balance, and accounts for its non-volatile fraction. An unsteady intermittent empirical model of droplet transport within the humid buoyant breath cloud, a random walk approach to address the droplet turbulent dispersion, and a droplet injection randomisation method complete the analysis. The model was validated against a set of CFD simulations and the little experimental data available in literature finding good agreement. In a second moment the breath cloud empirical model was abandoned in favour of a multi-scale approach where a set of unsteady CFD simulations were used to provide the local boundary conditions needed by the analytical solver to compute the droplet trajectory and evaporation history [19]. While the multi-scale approach provided slightly better results, the comparison confirmed the reasonable accuracy of the previous analytical/empirical. The approach proposed has been shown to be much cheaper in terms of computational resources needed with respect to more elaborated Eulerian-Lagrangian CFD analyses. Thanks to this, a large number of droplets can be easily simulated and the model has been used by the authors to perform statistical analyses of the droplets trajectory extrapolating spatial risk maps for different scenarios. To do so, assumptions on the droplet quantity and size distribution [8], on the viral concentration in the saliva [20], and on the virus titer decay [21] are necessary. The analyses were performed for a reference thermo-hygrometric condition of 20°C and 50% relative humidity. In the present work, the previous analysis is expanded by using the mentioned analytical/empirical model to

investigate the role of temperature and relative humidity on the exhaled droplet spatial distribution and on the critical droplet diameter. Different scenarios are compared for the case of coughing, namely: 10°C, 20°C, 30°C temperature, and 20%, 50%, 80% relative humidity.

2. ANALYTICAL MODEL

The analytical and the empirical models are briefly recalled in this section, for a more detailed analysis the reader is referred to the previous works from the authors [15, 18].

2.1 Trajectory Weight $\mathcal{W}$ and drag $\mathcal{D}$ are the main forces acting on the droplet during its flight

$$\mathcal{W} = \frac{1}{6} (\rho_p - \rho_m) \pi g D_p^3 \quad (1)$$

$$\mathcal{D} = -\frac{1}{8} C_d \rho_m \pi \|\mathbf{u}_r\| \mathbf{u}_r D_p^2 \quad (2)$$

where subscript p denotes the droplet, m the humid air mixture, ρ is the density, $\mathbf{g}$ the gravity acceleration vector, D the diameter, C_d the drag coefficient, and $\mathbf{u}_r = \mathbf{u}_p - \mathbf{u}_m$ the relative velocity vector. By integration, Newton's second law of motion provides equations for the droplet velocity $\mathbf{u}_p(t)$ and position $\mathbf{s}_p(t)$ in time, given the initial conditions $\mathbf{u}_{r,0}$ and $\mathbf{s}_{p,0}$

$$\mathbf{u}_p(t) = \mathbf{u}_m + \mathbf{u}_{r,0} \exp\left(-\frac{t}{\tau_d}\right) + \mathbf{u}_{r,t} \frac{\tau_d}{\tau_{d,t}} \left[1 - \exp\left(-\frac{t}{\tau_d}\right)\right] \quad (3)$$

$$\mathbf{s}_p(t) = \mathbf{s}_{p,0} + \left(\mathbf{u}_m + \mathbf{u}_{r,t} \frac{\tau_d}{\tau_{d,t}}\right) t + \left(\mathbf{u}_{r,0} - \mathbf{u}_{r,t} \frac{\tau_d}{\tau_{d,t}}\right) \tau_d \left[1 - \exp\left(-\frac{t}{\tau_d}\right)\right] \quad (4)$$

where the local velocity of the humid air $\mathbf{u}_m$ can be modelled at will to simulate different ventilation scenarios.

$$\tau_d = \frac{4\rho_p D_p}{3C_d \rho_m \|\mathbf{u}_r\|} \quad (5)$$

is the drag characteristic time and the subscript t denotes the terminal condition attained in the limit when $\dot{\mathbf{u}}_r = 0$. Drag coefficient is computed from the correlations for the flow past a sphere and, due to the small droplet size, must include Cunningham slip correction factor.

2.2 Evaporation Convection-diffusion equation in stationary form can be manipulated to provide a simple formula expressing the rate of change of the droplet diameter due to evaporation as

$$\dot{D}_p = -\frac{K}{2D_p} \quad (6)$$

where the evaporation rate $K = -dD_p^2/dt$ depends on many parameters such as the droplet density and temperature T_p , the binary diffusion coefficient of water in air D_v , Sherwood number, the water vapour gas constant R_v , and the vapour partial pressure difference between ambient and surface of the droplet ($p_{v,s} - p_{v,\infty}$)

$$K = \frac{4D_v \text{Sh}}{\rho_p R_v T_p} \frac{p}{p_{a,s}} (p_{v,s} - p_{v,\infty}) \quad (7)$$

By integration

$$D_p(t) = \sqrt{D_{p,0}^2 - Kt} \quad (8)$$

2.3 Energy Conservation of energy applied to the droplet requires that the sensible energy rate of change is balanced by convective heating and evaporative cooling. The thermal energy balance can be casted in terms of the droplet temperature rate of change

$$\dot{T}_p = \frac{T_{p,t} - T_p}{\tau_e} \quad (9)$$

where

$$\tau_e = \frac{c_{p,p} \rho_p D_p^2}{6 \lambda_s \text{Nu}} \quad (10)$$

is the droplet characteristic evaporation time, function of the droplet density, diameter, and specific heat $c_{p,p}$, the air thermal conductivity by the droplet surface λ_s , and Nusselt number. The droplet terminal temperature $T_{p,t} = T_\infty - \Delta T_p$ is the temperature reached by the droplet as a consequence of evaporative cooling, where T_∞ is the ambient temperature, and ΔT_p is the cooling term that depends on the water latent heat of vaporisation, the droplet density, the evaporation rate, the air thermal conductivity, and the Nusselt number. By integration

$$T_p(t) = T_{p,0} \exp\left(-\frac{t}{\tau_e}\right) + T_{p,t} \left[1 - \exp\left(-\frac{t}{\tau_e}\right)\right] \quad (11)$$

2.4 Species Assuming the droplet is formed by water and a nonvolatile solute solid fraction, from Raoult's law the water vapour partial pressure at the surface is reduced proportionally to the water molar fraction in the droplet y_{lq} . As a consequence the evaporation rate K is reduced accordingly and evaporation will terminate once the vapour partial pressure at the droplet surface is in equilibrium with that in the ambient. The water molar fraction can be expressed as a function of the liquid (lq) and nonvolatile (sl) molar mass M , density, and volume fraction φ as

$$y_{lq} = \left(1 + \frac{M_{lq} \rho_{sl} \varphi_{sl}}{M_{sl} \rho_{lq} \varphi_{lq}}\right)^{-1} \quad (12)$$

The droplet initial solid fraction is assumed to be equal to 1% in volume.

3. EMPIRICAL MODEL

3.1 Exhalation model An exhalation model is needed to evaluate the droplet transport within the warm humid breath cloud. The model proposed stems from uniform steady jets theory and momentum conservation, expanded to account for non-uniformity, unsteadiness, and viscous dissipation. It is assumed that the breath cloud grows as a cone with the distance and is characterised by a constant entrainment coefficient β so that at a generic distance r' the cloud half width is $\beta r'$ whereas at the mouth $\beta=1$ and $r'=r'_0$. The cloud thermo-hygrometric properties are computed from mixing, while the jet velocity decreases with the distance.

A non-uniform velocity profile is imposed assuming a self-similar radial profile given by any normalised function of the non-dimensional distance from the jet centreline ζ . A bell-shaped Gaussian function is chosen here. Introducing the non-dimensional distance $\xi = r' \beta / r'_{mt}$

$$\frac{u(\xi, \zeta)}{u_0} = \frac{1}{\xi} \sqrt{\frac{\rho_0}{\rho(\xi)}} \exp(-\zeta^2) \quad (13)$$

where the subscript 0 denotes the mouth cross-section. To account for unsteadiness and the possible periodicity of respiratory events, it is assumed that the velocity profile shape of eq. (13) is retained in time and is scaled by a factor $\gamma(t)$ derived from momentum conservation, thus providing the final velocity field in the breath cloud

$$\frac{u(\xi, \zeta, t)}{u_0} = \frac{1}{\xi} \sqrt{\frac{\rho_0}{\rho(\xi)}} \exp(-\zeta^2) \gamma(t) \quad (14)$$

With the due considerations, the function $\gamma(t)$ can be derived for any type of unsteady and non-uniform velocity respiratory impulse, and can embed an exponential decay term to account for momentum dissipation.

The breath cloud buoyancy is enforced by applying an empirical model taken from the literature for steady jets [22]. The jet potential core is modelled by enforcing $\xi=1$ up to a certain distance from the mouth as suggested in the literature [16] and rescaling ξ accordingly at further distances.

In the present work, the coughing impulse is modelled as a piecewise positive sinusoidal function peaking at a generic time $t_{pk}=0.1$ s and terminating at time $t_{ex}=0.4$ s. The overall exhaled volume and the peak velocity are chosen as $V_{ex}=1$ dm³ and $u_{pk}=10$ m/s [23], implying a mouth opening diameter of ≈ 22 mm. The breath cloud is supposed to be expelled upward at an angle 27.5° with respect to the horizontal [24] from a height of 1.5 m, with a temperature of 34° C and relative humidity of 95%. The entrainment coefficient is set to $\beta=0.11$ [25].

3.2 Turbulent dispersion and droplet randomisation Turbulent dispersion is modelled using a discrete random walk approach. This is a standard approach in CFD. The velocity fluctuation $\mathbf{u}'$ is modelled as a random function of the turbulent kinetic energy which is updated periodically and added to the local air velocity $\mathbf{u}_m$ in eqs. (3) and (4).

It is assumed that droplets are randomly released from the mouth cross-section in a proportional way to the local velocity both in space and time. In this way a Gaussian spatial distribution injection is obtained where droplets are more likely to be released close to the velocity peak time.

These random elements in a model that would otherwise be deterministic allow statistical analyses over the droplet trajectories to be performed. Droplets are assumed to be released in a condition of thermal equilibrium with the breath cloud environment and with zero relative velocity.

3.3 Inhalation model To transform the droplet trajectory data into a probability of inhaling a certain amount of viruses and become infected, an inhalation model is needed.

Here, it is simply assumed that a breathing susceptible individual will inhale a hemispheric volume in front of his nose at each breath. Assuming a uniform inhalation over a given periodic time span, the inhaled air velocity $\mathbf{u}_{in}$ can be estimated as a function of the radial distance from the individual nose. A droplet crossing the inhaled volume is assumed to be inhaled, with a probability equal to the breathing inhalation time over the breathing period, if its velocity component orthogonal to $\mathbf{u}_{in}$ is smaller than $\|\mathbf{u}_{in}\|$.

The inhaled volume is assumed to be $V_{in}=600$ cm³ per breath, the breathing period equal to $t_{pr}=4$ s and the inhalation time $t_{in}=2.4$ s.

4. EXHALATION DATA

In addition to the foregoing fluid dynamic model, to quantify the risk of infectious disease transmission additional information is needed. The exhaled droplet size distribution is assumed to follow a power-law Probability Density Function (PDF) $\mathcal{B}$ in line with what suggested in [8]

$$\mathcal{B}(D_p) = \frac{B}{D_p^m} \quad (15)$$

with B and m to be determined. Since smaller droplets are deemed more dangerous being able to penetrate into the lower airways, a droplet diameter dependent infectious dose $\mathcal{I}$ power-law PDF is further assumed

$$\mathcal{I}(D_p) = ID_p^n \quad (16)$$

where I and n are constants to be determined. With $m=2$ and $n=1$ and integrating over a generic range of possible droplet diameters $[D_{\min}, D_{\max}] = [10^{-7} \text{ m}, 10^{-3} \text{ m}]$ [2], the number of exhaled droplets N_D , their cumulative volume V_D , and the number of quanta exhaled Q_D (assuming a uniform virus concentration c in the saliva) are evaluated as

$$N_D = \int_{D_{\min}}^{D_{\max}} \frac{B}{D_p^m} dD_p = B \left(\frac{1}{D_{\min}} - \frac{1}{D_{\max}} \right) \approx \frac{B}{D_{\min}} \quad (17)$$

$$V_D = \int_{D_{\min}}^{D_{\max}} \frac{B}{D_p^m} \frac{\pi D_p^3}{6} dD_p = \frac{\pi B}{12} (D_{\max}^2 - D_{\min}^2) \approx \frac{\pi B D_{\max}^2}{12} \quad (18)$$

$$Q_D = \int_{D_{\min}}^{D_{\max}} \frac{B}{D_p^m} \frac{\pi D_p^3}{6} \frac{c}{ID_p^n} dD_p = \frac{\pi B c}{6I} (D_{\max} - D_{\min}) \approx \frac{\pi B c D_{\max}}{6I} \quad (19)$$

From the definition of average infectious dose as the quantity of viruses corresponding to one quanta $\bar{I} = cV_D/Q_D$, where one quanta is the virus dose causing the disease with a probability of $1 - \exp(-1) = 63\%$ when inhaled

$$I = \frac{2\bar{I}}{D_{\max} + D_{\min}} \approx \frac{2\bar{I}}{D_{\max}} \quad (20)$$

$$Q_D = \frac{\pi B c}{12\bar{I}} (D_{\max}^2 - D_{\min}^2) \approx \frac{\pi B c D_{\max}^2}{12\bar{I}} \quad (21)$$

In the following, the virus concentration in the saliva is assumed to be $c = 3300 \text{ copies/mm}^3$. This corresponds to an average value reported in the literature for SARS CoV-2 virus [20]. The average infectious dose is believed to be $\bar{I} \approx 100 \text{ copies}$ for many infectious diseases [26]. Last, the amount of saliva expelled during a cough is estimated to be $V_D \approx 5 \text{ mm}^3$, in line with [27], resulting in $B = 20 \text{ mm}$. It follows that the number of droplets exhaled and the number of quanta associated to coughing are $N_D = 2 \times 10^5$ and $Q_D = 172.8$ quanta, respectively.

5. SIMULATION CAMPAIGN

During the simulation campaign, atmospheric pressure is assumed in a quiescent environment. Nine scenarios are investigated involving different ambient temperatures and relative humidities, namely: 10°C , 20°C , 30°C , and 20%, 50%, 80%. In each scenario a set of 972 logarithmically distributed diameters are tested. The range from $0.1 \mu\text{m}$ to 1 mm is subdivided in 972 parts and the median diameter in each sub-range is picked. 21

droplets are simulated for each diameter resulting in 20412 droplets overall. A number of quanta obtained from eq. (21) is associated to each range. In this way, even though the size distribution simulated does not reflect that in eq. (15), the viral distribution does and the trajectories collected cover uniformly the range of interest.

Droplets are tracked up to when they either deposit on the ground or terminate evaporation reaching thermal equilibrium with the environment having a terminal velocity $< 2 \text{ cm/s}$ outside of the breath cloud volume, in which case they are assumed airborne. For each droplet, a large set of information is collected at each time step, including position, velocity, diameter, temperature, local air velocity and thermo-hygrometric conditions.

Each scenario required on average 7h time to be completed running on a small cluster using 21 CPUs. This corresponds to $\approx 25 \text{ s}$ of CPU time per droplet. Nonetheless, the time required by the simulation of a single droplet varies much depending on its size and ranges from just a few seconds for large droplets that deposit quickly, up to a couple of minutes for small droplets remaining trapped in the breath cloud for long.

6. POST-PROCESSING

Let us consider the exhaled droplets that in their journey cross a generic volume of space V_c and let the residence time of the droplet inside the volume and its viral load be t_i and Q_i , respectively. The *viral exposure* of the volume [28] is defined as

$$e_c = \frac{1}{V_c} \sum_i Q_i t_i \quad (22)$$

The number of quanta inhaled, along with the volume V_c by a susceptible individual breathing normally is

$$i_c = \frac{t_{\text{in}}}{t_{\text{pr}}} \sum_i Q_i \quad (23)$$

where the sum is made on all droplets intercepting the volume that comply with the inhalation condition in §3.3. From the definition of quanta, the risk to contract the disease once a dose i_c has been inhaled is

$$r_c = 1 - \exp(-i_c) \quad (24)$$

Equations (22)-(24) account for direct inhalation route. Airborne route can be included, under the assumption of well-mixed airborne fraction, using Wells-Riley model [29]. Coughing suddenly introduces an amount of airborne quanta Q_{ab} in an environment of volume V_{rm} . If τ_{rm} is the characteristic time associated to viral removal rate, e.g. by titer decay and/or ventilation, the viral concentration in the room c_{rm} decays as

$$c_{\text{rm}}(t) = \frac{Q_{\text{ab}}}{V_{\text{rm}}} \exp\left(-\frac{t}{\tau_{\text{rm}}}\right) \quad (25)$$

or, on a time-average basis

$$\bar{c}_{\text{rm}}(t) = \frac{Q_{\text{ab}}}{V_{\text{rm}}} \frac{\tau_{\text{rm}}}{t} \left[1 - \exp\left(-\frac{t}{\tau_{\text{rm}}}\right) \right] \quad (26)$$

Including airborne contribution, eqs. (22)-(24) can be rewritten as

$$\hat{e}_c = \frac{1}{V_c} \sum_i Q_i t_i + \bar{c}_{rm}(t) t \quad (27)$$

$$\hat{i}_c = \frac{t_{in}}{t_{pr}} \sum_i Q_i + \bar{c}_{rm}(t) V_{in} \frac{t}{t_{pr}} \quad (28)$$

$$\hat{r}_c = 1 - \exp(-\hat{i}_c) \quad (29)$$

where t/t_{pr} is the number of breaths during which the susceptible individual remains exposed to the airborne viral concentration. Wells-Riley model can also be used to evaluate the time-varying viral concentration on the ground associated to fomite transmission risk due to the quanta Q_{gr} deposited on a surface S_{gr} after a cough

$$c_c = \frac{Q_{gr}}{S_{gr}} \exp\left(-\frac{t}{\tau_{gr}}\right) \quad (30)$$

In the following, the time-dependent airborne and fomite components are evaluated for an exposure time of 1 h after a single cough in a room of $V_{rm}=300\text{ m}^3$ assuming $\tau_{rm}=45\text{ min}$ and $\tau_{gr}=9\text{ h}$ [21] starting from a virus-free environment at time zero.

During post-processing, the volume and the ground in front of the infected individual are discretised with cells of 4 mm. For each cell eqs. (27)-(30) are evaluated by processing the trajectories of all simulated droplets. As a result, spatial maps of viral exposure $\hat{e}_c$, inhaled quanta $\hat{i}_c$, inhalation transmission risk $\hat{r}_c$, and viral concentration on the ground c_c are made available. For visualization purposes, the spatial three-dimensional maps are projected on the exhalation midplane where data is averaged over the local breath cloud width.

7. RESULTS

Table 1 and Fig. 1 show how the critical diameter and the number of airborne quanta released after a cough vary with the temperature and the ambient relative humidity. Of course, the warmer and the drier the environment the most evaporation is favoured, leading to large critical diameters, large amounts of airborne quanta, and an increased airborne transmission risk. On the contrary, cold and humid environments hamper evaporation and favour the droplet deposition on the ground, thus boosting fomite transmission. For the $T=30^\circ\text{C}$, $\text{RH}=20\%$ scenario the critical diameter and the number of airborne quanta are almost double compared to the $T=10^\circ\text{C}$, $\text{RH}=80\%$ one, and so is the risk of airborne transmission. Nonetheless, it must be noted that the difference among these scenarios is ≈ 8.5 quanta, which is just a small fraction ($\approx 5\%$) of the 172.8 quanta emitted with a cough according to the model outlined in §4. This suggests that while airborne transmission will be largely affected by the ambient thermo-hygrometric conditions, direct inhalation and fomite transmission will not.

For the sake of brevity, in the following only the results of these two limit scenarios are shown and compared.

Table 1 Critical diameter and airborne quanta for the different scenarios investigated.

Relative Humidity	Temperature		
	10 °C	20 °C	30 °C
20%	$D_{cr}=91.6\text{ }\mu\text{m}$ $Q_{ab}=15.81$	$D_{cr}=102.1\text{ }\mu\text{m}$ $Q_{ab}=17.62$	$D_{cr}=108.9\text{ }\mu\text{m}$ $Q_{ab}=18.80$
50%	$D_{cr}=78.8\text{ }\mu\text{m}$ $Q_{ab}=13.60$	$D_{cr}=87.3\text{ }\mu\text{m}$ $Q_{ab}=15.07$	$D_{cr}=94.9\text{ }\mu\text{m}$ $Q_{ab}=16.38$
80%	$D_{cr}=59.5\text{ }\mu\text{m}$ $Q_{ab}=10.27$	$D_{cr}=66.0\text{ }\mu\text{m}$ $Q_{ab}=11.38$	$D_{cr}=70.8\text{ }\mu\text{m}$ $Q_{ab}=12.21$

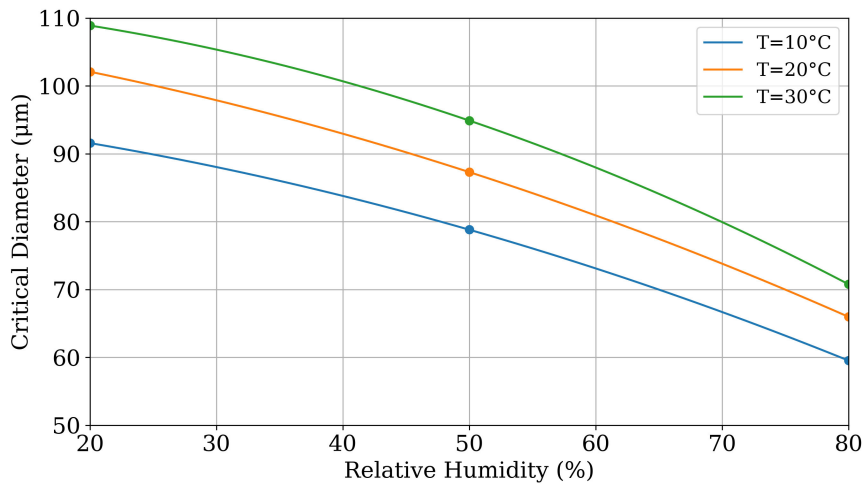


Fig. 1 Critical diameter for the different scenarios investigated.

Figure 2 shows the droplet trajectories coloured by their initial size. Three different behaviours are highlighted: large droplets follow a quasi-parabolic trajectory being little affected by drag, small droplets evaporate almost immediately and remain trapped within the breath cloud due to their small terminal velocity, intermediate size droplets are able to exit the breath cloud influence area by gravity at some point during their journey and, being the ambient quiescent, fall vertically as their horizontal velocity is rapidly damped by drag forces. Figure 3 shows the same information limited to airborne droplet trajectories above $10\mu\text{m}$ in size. The breath cloud is shown to remain coherent up to a distance of $\approx 1\text{m}$ from the coughing individual. The hot and dry scenario of course is characterised by a larger number of airborne droplets and also by a larger number of droplets that remain within the breath cloud, while in the cold and humid scenario the role of buoyancy in deviating the breath cloud upward can be perceived.

Figure 4 shows the viral concentration on the ground 1 h after the respiratory event for the two scenarios. As expected there is no apparent difference among the two cases for what concerns fomite transmission. Larger droplets can deposit on the ground up to 2 m away from the infected individual, and viral concentrations up to 0.3 quanta/cm^2 are found, meaning that the amount of viruses that could be collected by a simple fingertip touch are likely to cause the disease if brought in contact with the airways and inhaled.

To conclude, Figs. 5-7 show the spatial distribution maps of the viral exposure $\hat{e}_c$, of the viral inhalation $\hat{i}_c$, and of the risk of infection $\hat{r}_c$ as introduced in §6. These quantities are derived from one another, and of course are largely correlated. The traces of the droplet trajectories are mostly linked to direct inhalation transmission risk, while the map background colour to the airborne transmission risk after 1 h exposure. As such, the main difference that may be noted among the two scenarios in these images is, in fact, in the lighter background colour that clearly highlights the increased risk of airborne transmission associated to the hot and dry scenario.

Airborne contribution to the viral exposure amounts to $1.25 \times 10^{-4}\text{ quanta} \cdot \text{s/m}^3$ in the hot and dry scenario, and to $6.81 \times 10^{-5}\text{ quanta} \cdot \text{s/m}^3$ in the cold and wet one. This translates into the inhalation of $1.87 \times 10^{-2}\text{ quanta}$ and $1.02 \times 10^{-2}\text{ quanta}$ after just 1 h of exposure, respectively, corresponding to infection risks of 1.85% and 1.02%. Although these risks of infection are apparently low, airborne transmission becomes a real issue in case of repeated and prolonged exposures in relatively crowded environments such as it occurs in offices and school classrooms. Further, airborne transmission is particularly challenging to mitigate as conventional preventive measures that can be taken by individuals, such as maintaining physical distance, do not ensure protection.

Small and intermediate droplet trajectories, form a compact high virus density area covering the space between the infected individual mouth and the ground within a 1 m distance. As such, 1 m interpersonal distance is barely sufficient in both cases to protect the susceptible individual from direct inhalation of small droplets from the breath cloud, but is not enough to protect from direct inhalation of large droplets that may occur if the susceptible individual is at a lower height compared to the infected one. This may occur, for instance, when the infected individual is standing in front of a desk where the susceptible individual is sitting. Further, such a condition is also very dangerous in view of surface infection and fomite transmission through the desk surface.

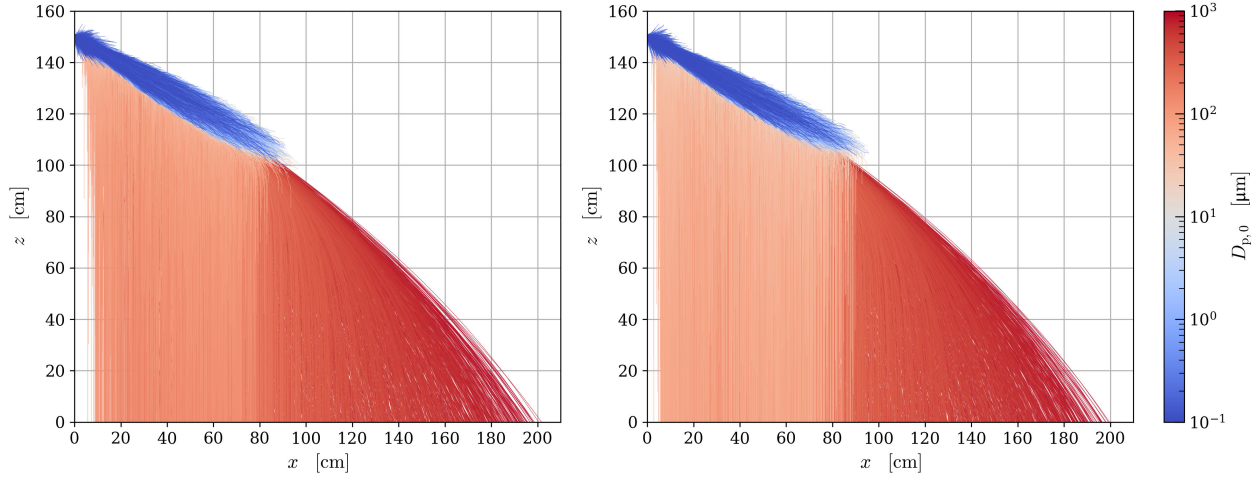


Fig. 2 Droplet trajectories coloured by their initial diameter: $T=30^{\circ}\text{C}$, $\text{RH}=20\%$ scenario (left), and $T=10^{\circ}\text{C}$, $\text{RH}=80\%$ scenario (right).

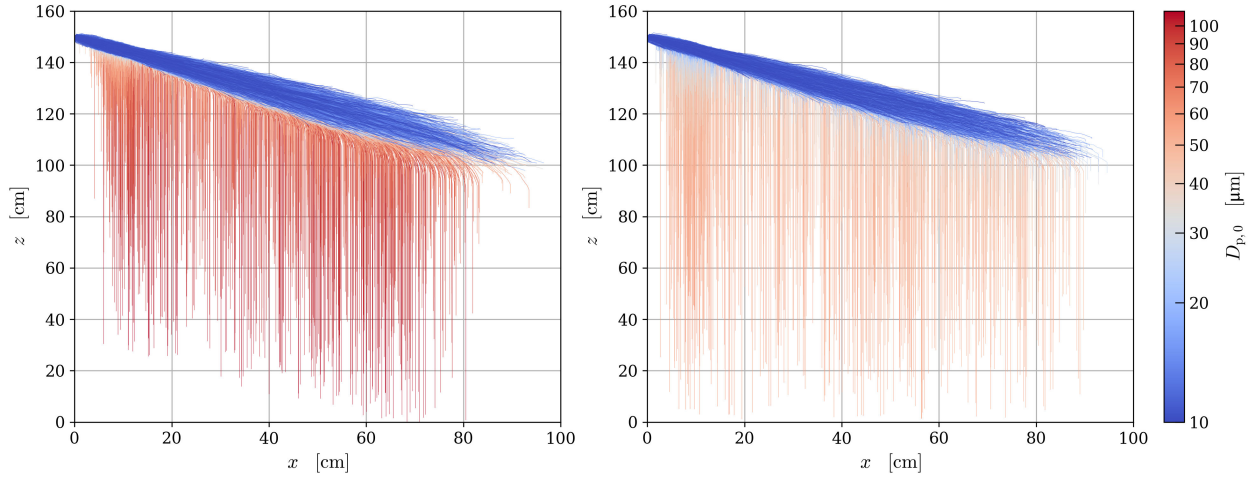


Fig. 3 Airborne droplet trajectories coloured by their initial diameter: $T=30^{\circ}\text{C}$, $\text{RH}=20\%$ scenario (left), and $T=10^{\circ}\text{C}$, $\text{RH}=80\%$ scenario (right).

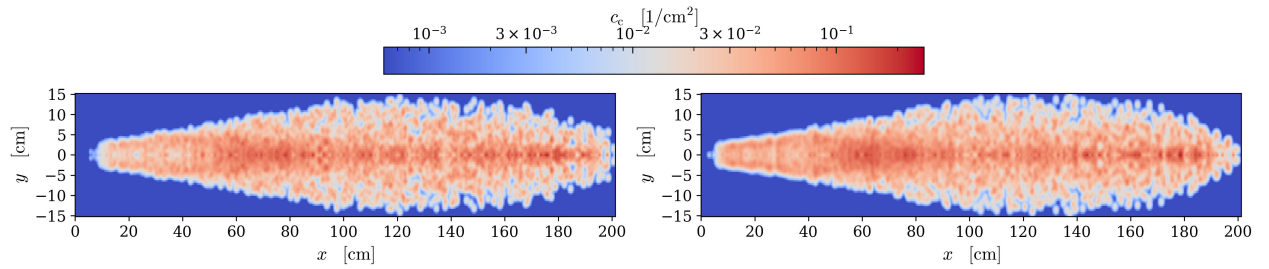


Fig. 4 Viral concentration on the ground expressed in quanta/ cm^2 : $T=30^{\circ}\text{C}$, $\text{RH}=20\%$ scenario (left), and $T=10^{\circ}\text{C}$, $\text{RH}=80\%$ scenario (right).

8. CONCLUSIONS

The risk of infectious disease transmission under different scenarios has been evaluated, from a purely fluid dynamics perspective, using an analytical model recently developed by the authors that simulates the transport and the evaporation of respiratory droplets. The analytical tool is coupled to a series of empirical models

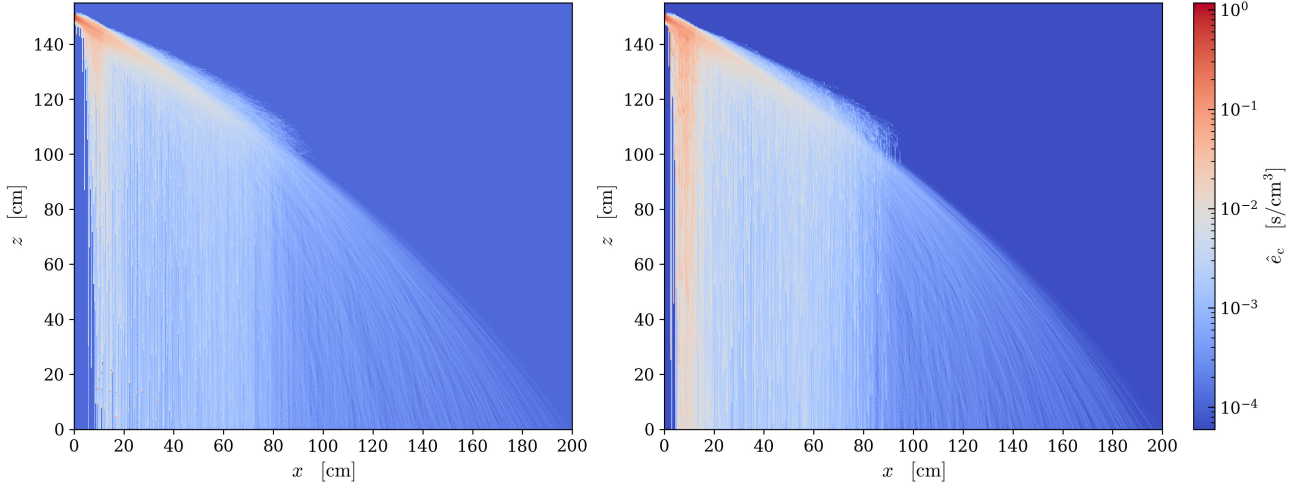


Fig. 5 Viral exposure map expressed in $\text{quanta} \cdot \text{s} / \text{cm}^3$: $T=30^\circ\text{C}$, $\text{RH}=20\%$ scenario (left), and $T=10^\circ\text{C}$, $\text{RH}=80\%$ scenario (right).

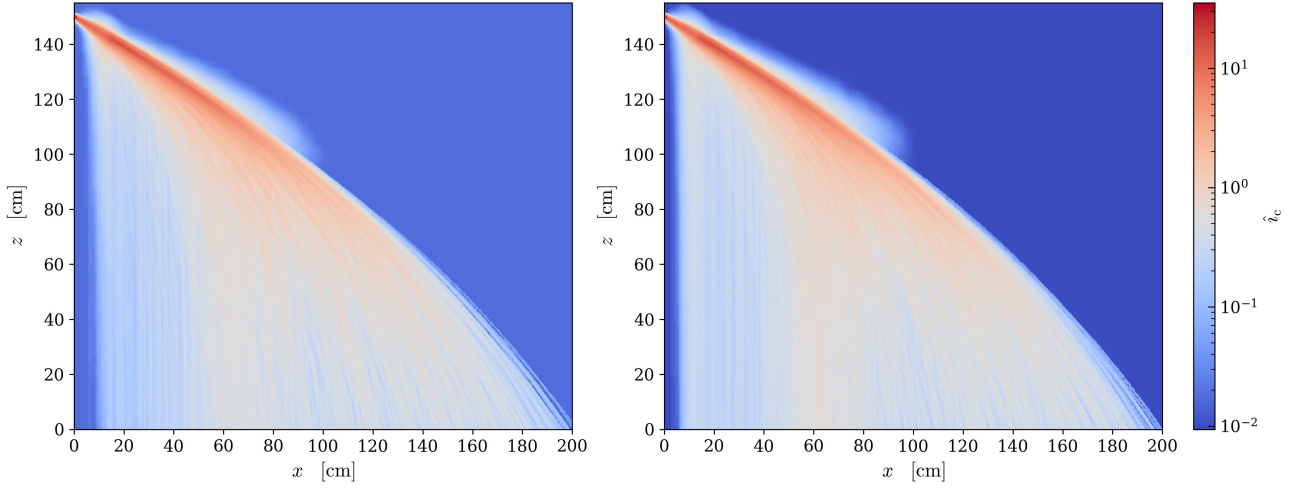


Fig. 6 Viral inhalation map expressed in quanta: $T=30^\circ\text{C}$, $\text{RH}=20\%$ scenario (left), and $T=10^\circ\text{C}$, $\text{RH}=80\%$ scenario (right).

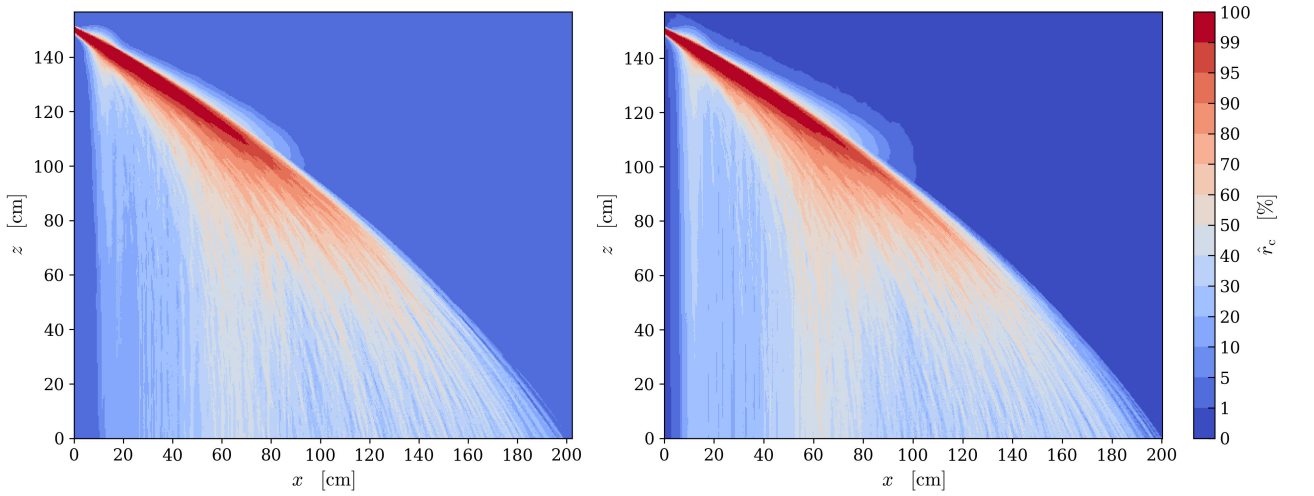


Fig. 7 Risk of infection map: $T=30^\circ\text{C}$, $\text{RH}=20\%$ scenario (left), and $T=10^\circ\text{C}$, $\text{RH}=80\%$ scenario (right).

to account for the droplet transport within the humid air breath cloud, the random walk droplet turbulent dispersion, and the randomised droplet exhalation. These stochastic features allow statistical analyses on the droplet trajectory distribution to be performed over large sets of droplets. With assumptions on the droplet size distribution and viral content, the trajectory distribution has been transformed into spatial maps of viral exposure, viral inhalability, viral concentration, and risk of infection accounting for direct inhalation, airborne, and fomite transmission risks.

Nine scenarios were investigated simulating more than 20 thousand droplets in each run. The scenarios were chosen to assess the role of thermo-hygrometric conditions on the risk of infection. It was found that the ambient temperature and relative humidity can have a significant impact on the risk of airborne transmission, risk that is almost doubled passing from a reference cold and wet condition ($T=10^{\circ}\text{C}$, $\text{RH}=80\%$) to a hot and dry one ($T=30^{\circ}\text{C}$, $\text{RH}=20\%$). On the contrary, direct inhalation and fomite transmission are essentially unaffected by the ambient thermo-hygrometric conditions.

NOMENCLATURE

$\mathcal{B}$	droplet size PDF	(m^{-1})	ζ	centreline dimensionless distance	$(-)$
C_d	drag coefficient	$(-)$	ξ	radial dimensionless distance	$(-)$
c	concentration	(m^{-3})	ρ	density	(kg/m^3)
c_p	specific heat at constant pressure	(J/kgK)	τ	characteristic time	(s)
D	diameter	(m)	φ	volume fraction	$(-)$
D_v	binary diffusion coefficient	(m^2/s)	<i>subscripts</i>		
$\mathcal{D}$	drag force	(N)	a	dry air	
e	viral exposure	(s/m^3)	ab	airborne	
$\mathbf{g}$	gravitational acceleration	(m/s)	c	cell	
$\mathcal{I}$	infectious dose PDF	(m^{-1})	cr	critical	
i	viral inhalation	$(-)$	D	exhaled droplets	
K	evaporation rate	(m^2/s)	d	drag	
M	molar mass	(kg/kmol)	e	evaporation	
N	number	$(-)$	ex	exhalation	
Nu	Nusselt number	$(-)$	gr	ground	
p	pressure	(Pa)	in	inhalation	
Q	exhaled quanta	$(-)$	lq	liquid fraction	
R	Specific gas constant	(J/kgK)	m	humid air mixture	
r	risk of infection	$(-)$	max	maximum	
r'	radius or radial distance	(m)	min	minimum	
S	surface area	(m^2)	mt	mouth or nose orifice	
$\mathbf{s}$	position	(m)	p	droplet or particle	
Sh	Sherwood number	$(-)$	pk	peak	
T	temperature	(K)	pr	period	
t	time	(s)	r	relative	
$\mathbf{u}$	velocity	(m/s)	rm	room	
V	volume	(m^3)	s	air value by the droplet surface	
$\mathcal{W}$	weight force	(N)	sl	solid fraction	
y	molar fraction	$(-)$	t	terminal condition	
<i>greek symbols</i>			v	water vapour	
β	entrainment coefficient	$(-)$	0	initial condition	
γ	cloud velocity scaling factor	$(-)$	∞	ambient condition	

REFERENCES

- [1] Leung, N. H. L. and Milton, D. K., “New WHO proposed terminology for respiratory pathogen transmission”, *Nature Reviews Microbiology*, 22(8), pp. 453–454 (2024).

- [2] Bourouiba, L., “Fluid dynamics of respiratory infectious diseases”, *Annual Review of Biomedical Engineering*, 23, pp. 547–577 (2021).
- [3] Xie, X., Li, Y., Chwang, A. T. Y., Ho, P. L. and Seto, W. H., “How far droplets can move in indoor environments – revisiting the Wells evaporation–falling curve”, *Indoor Air*, 17(3), pp. 211–225 (2007).
- [4] Xu, C., Nielsen, P. V., Liu, L., Jensen, R. L. and Gong, G., “Human exhalation characterization with the aid of schlieren imaging technique”, *Building and Environment*, 112, pp. 190–199 (2017).
- [5] Tang, J. W., Nicolle, A. D., Klettner, C. A., Pantelic, J., Wang, L., Suhaimi, A. B., Tan, A. Y. L., Ong, G. W. X., Su, R., Sekhar, C., Cheong, D. D. W. and Tham, K. W., “Airflow dynamics of human jets: sneezing and breathing – potential sources of infectious aerosols”, *Plos One*, 8(4), p. e59970 (2013).
- [6] Alsved, M., Matamis, A., Bohlin, R., Richter, M., Bengtsson, P.-E., Fraenkel, C.-J., Medstrand, P. and Löndahl, J., “Exhaled respiratory particles during singing and talking”, *Aerosol Science and Technology*, 54(11), pp. 1245–1248 (2020).
- [7] Asadi, S., Wexler, A. S., Cappa, C. D., Barreda, S., Bouvier, N. M. and Ristenpart, W. D., “Aerosol emission and superemission during human speech increase with voice loudness”, *Scientific Reports*, 9, p. 2348 (2019).
- [8] Balachandar, S., Zaleski, S., Soldati, A., Ahmadi, G. and Bourouiba, L., “Host-to-host airborne transmission as a multiphase flow problem for science-based social distance guidelines”, *International Journal of Multiphase Flow*, 132, p. 103439 (2020).
- [9] Motamedi, H., Shirzadi, M., Tominaga, Y. and Mirzaei, P., “CFD modeling of airborne pathogen transmission of COVID-19 in confined spaces under different ventilation strategies”, *Sustainable Cities and Society*, 76, p. 103397, doi:10.1016/j.scs.2021.103397 (2022).
- [10] Vuorinen, V., Aarnio, M. A., Alava, M., Alopaeus, V., Atanasova, N., Auvinen, M., Balasubramanian, N., Bordbar, H., Erästö, P., Grande, R., Hayward, N., Hellsten, A., Hostikka, S., Hokkanen, J., Kaario, O., Karvinen, A., Kivistö, I., Korhonen, M., Kosonen, R., Kuusela, J., Lestinen, S., Laurila, E., Nieminen, H. J., Peltonen, P., Pokki, J., Puisto, A., Råback, P., Salmenjoki, H., Sironen, T. and Österberg, M., “Modelling aerosol transport and virus exposure with numerical simulations in relation to SARS-CoV-2 transmission by inhalation indoors”, *Safety Science*, 130, p. 104866 (2020).
- [11] Arpino, F., Cortellessa, G., Grossi, G. and Nagano, H., “A Eulerian-Lagrangian approach for the non-isothermal and transient CFD analysis of the aerosol airborne dispersion in a car cabin”, *Building and Environment*, 209, p. 108648 (2022).
- [12] Chao, C., Wan, M., Morawska, L., Johnson, G., Ristovski, Z., Hargreaves, M., Mengersen, K., Corbett, S., Li, Y., Xie, X. and Katoshevski, D., “Characterization of expiration air jets and droplet size distributions immediately at the mouth opening”, *Journal of Aerosol Science*, 40, pp. 122–133 (2009).
- [13] Parienta, D., Morawska, L., Johnson, G. R., Ristovski, Z. D., Hargreaves, M., Mengersen, K., Corbett, S., Chao, C. Y. H., Li, Y. and Katoshevski, D., “Theoretical analysis of the motion and evaporation of exhaled respiratory droplet of mixed composition”, *Journal of Aerosol Science*, 42(1), pp. 1–10 (2011).
- [14] Bourouiba, L., Dehandschoewercker, E. and Bush, J. W. M., “Violent expiratory events: on coughing and sneezing”, *Journal of Fluid Mechanics*, 745, pp. 537–563 (2014).
- [15] Cavazzuti, M. and Tartarini, P., “Statistical analysis of infectious disease transmission risk based on exhaled respiratory droplet trajectory distribution”, *Physics of Fluids*, 36(6), p. 063341 (2024).
- [16] Chen, W., Zhang, N., Wei, J., Yen, H.-L. and Li, W., “Short-range airborne route dominates exposure of respiratory infection during close contact”, *Building and Environment*, 176, p. 106859 (2020).
- [17] Redrow, J., Mao, S., Celik, I., Posada, J. A. and Feng, Z.-G., “Modeling the evaporation and dispersion of airborne sputum droplets expelled from a human cough”, *Building and Environment*, 46(10), pp. 2042–2051 (2011).
- [18] Cavazzuti, M. and Tartarini, P., “Transport and evaporation of respiratory droplets: an analytical model”, *Physics of Fluids*, 35(10), p. 103327 (2023).
- [19] Cavazzuti, M., Campanelli, L. and Tartarini, P., “Estimating infectious disease transmission risk: an analysis based on multi-scale modelling of respiratory droplet transport”, *Physics of Fluids*, 37(2), p. 023317 (2025).
- [20] To, K. K.-W., Tsang, O. T.-Y., Yip, C. C.-Y., Chan, K.-H., Wu, T.-C., Chan, J. M.-C., Leung, W.-S., Chik, T. S.-H., Choi, C. Y.-C., Kandamby, D. H., Lung, D. C., Tam, A. R., Poon, R. W.-S., Fung, A. Y.-F., Hung, I. F.-N., Cheng, V. C.-C., Chan, J. F.-W. and Yuen, K.-Y., “Consistent detection of 2019 novel coronavirus in saliva”, *Clinical Infectious Diseases*, 71(15), pp. 841–843 (2020).
- [21] van Doremalen, N., Bushmaker, T., Morris, D. H., Holbrook, M. G., Gamble, A., Williamson, B. N., Tamin, A., Harcourt, J. L., Thornburg, N. J., Gerber, S. I., Lloyd-Smith, J. O., de Wit, E. and Munster, V. J., “Aerosol and surface stability of SARS-CoV-2 as compared with SARS-CoV-1”, *New England Journal of Medicine*, 382(16), pp. 1564–1567 (2020).
- [22] Baturin, V. V., *Fundamentals of industrial ventilation*, Oxford: Pergamon Press (1972).
- [23] Li, H., Leong, F. Y., Xu, G., Ge, Z., Kang, C. W. and Lim, K. H., “Dispersion of evaporating cough droplets in tropical outdoor environment”, *Physics of Fluids*, 32(11), p. 113301 (2020).
- [24] Busco, G., Yang, S. R., Seo, J. and Hassan, Y. A., “Sneezing and asymptomatic virus transmission”, *Physics of Fluids*, 32(7), p. 073309 (2020).
- [25] Gupta, J. K., Lin, C.-H. and Chen, Q., “Flow dynamics and characterization of a cough”, *Indoor Air*, 19(6), pp. 517–525 (2009).
- [26] Kolinski, J. M. and Schneider, T. M., “Superspreading events suggest aerosol transmission of SARS-CoV-2 by accumulation in enclosed spaces”, *Physical Review E*, 103(3), p. 033109 (2021).
- [27] Bahl, P., Doolan, C., de Silva, C. A. A., C., Bourouiba, L. and MacIntyre, C. R., “Airborne or droplet precautions for health workers treating coronavirus disease 2019?”, *Journal of Infectious Diseases*, 225(9), pp. 1561–1568 (2020).
- [28] Foat, T. G., Higgins, B., Abbs, C., Maishman, T., Coldrick, S., Kelsey, A., Iivings, M. J., Parker, S. T. and Noakes, C. J., “Modelling the effect of temperature and relative humidity on exposure to SARS-CoV-2 in a mechanically ventilated room”, *Indoor Air*, 32(11), p. e13146 (2022).
- [29] Riley, E. C., Murphy, G. and Riley, R. L., “Airborne spread of measles in a suburban elementary school”, *American Journal of Epidemiology*, 107(5), pp. 421–432 (1978).