Since January 2020 Elsevier has created a COVID-19 resource centre with free information in English and Mandarin on the novel coronavirus COVID-19. The COVID-19 resource centre is hosted on Elsevier Connect, the company's public news and information website.

Elsevier hereby grants permission to make all its COVID-19-related research that is available on the COVID-19 resource centre - including this research content - immediately available in PubMed Central and other publicly funded repositories, such as the WHO COVID database with rights for unrestricted research re-use and analyses in any form or by any means with acknowledgement of the original source. These permissions are granted for free by Elsevier for as long as the COVID-19 resource centre remains active.
Development and evaluation of a fluidic facemask for airborne transmission mitigation

David Keisar a, Anan Garzozi a, Moshe Shoham b, David Greenblatt b, * 

a Grand Technion Energy Program (GTEP), Technion – Israel Institute of Technology, Technion Campus, Haifa 3200003, Israel  
b Faculty of Mechanical Engineering, Technion – Israel Institute of Technology, Technion Campus, Haifa 3200003, Israel

ABSTRACT

Recently, a fluidic facemask concept was proposed to mitigate the transmission of virus-laden aerosol and droplet infections, such as SARS-CoV-2 (COVID-19). This paper describes an experimental investigation of the first practical fluidic facemask prototype, or “Air-Screen”. It employs a small, high-aspect-ratio, crossflow fan mounted on the visor of a filter-covered cap to produce a rectangular air jet, or screen, in front of the wearer’s face. The entire assembly weighs less than 200 g. Qualitative flow visualization experiments using a mannequin clearly illustrated the Air-Screen’s ability to effectively block airborne droplets (~10 μm) from the wearer’s face. Quantitative experiments to simulate droplets produced during sneezing or a wet cough (~10 μm) were propelled (via a transmitter) at an average velocity of 50 m/s at 1 m from the mannequin or a target. The Air-Screen blocked 62% of all droplets with a diameter of less than 150 μm. A mathematical model, based on a weakly-advected jet in a crossflow, was employed to gain greater insight into the experimental results. This investigation highlighted the remarkable blocking effect of the Air-Screen and serves as a basis for a more detailed and comprehensive experimental evaluation.

1. Introduction

The most pervasive and visible consequence of the recent SARS-CoV-2 (COVID-19) pandemic is the wearing of surgical or cotton facemasks for the purpose of mitigating human-to-human transmission of the virus. The U.S. Centers for Disease Control and Prevention (CDC) determined that an infected or asymptomatic individual wearing a surgical mask protects others from the wearer [1]. This is because the primary mechanism of transmission between individuals is considered to be virus-laden aerosols and droplets [2]. N95 masks are considered to be more effective than common disposable masks [3] and are approved by the US National Institute for Occupational Safety and Health (NIOSH). Despite extensive research [4–6], it is still unclear how effective masks are in protecting the wearer from being infected by others [7]. In early 2020, the U.S. Food and Drug Administration (FDA) had not approved any surgical mask specifically for protection against COVID-19 [8]. However, later that year, the FDA issued an umbrella Emergency Use Authorizations (EUA) for certain disposable, single-use surgical masks. Even when wearing surgical masks, isolated droplets can still be spread beyond 70 cm [9] or even greater distances when using bandanas or N95 masks [10,11]. Likewise, in the presence of wind towards the wearer, as in open-air environments, large droplets (up to 80 μm) can be carried up to 6 m downwind [12]. The distinction between aerosols and droplets, traditionally considered to be less than 5 μm to 10 μm, has important implications for transmission. For example, although 10 μm droplets are not considered aerosols, they remain airborne for approximately 500 s in the absence of evaporation and are sometimes denoted as “airborne droplets” [13]. The effectiveness of surgical and other facemasks is usually determined for virus-laden droplets with diameters greater than D = 5 μm [14], usually generated by sneezing and coughing [2,15]. However, smaller-sized virus-laden aerosols with diameters less than 5 μm, [14,16] are considered a primary transmission source for other viral

Abbreviations: CDC, Centers for Disease Control and Prevention; ClO₂, Chlorine dioxide; DEHS, DiEthyl-Hexyl-Sebacate; EUA, Emergency Use Authorizations; HEPA, High-efficiency particulate air; IGV, Inlet guide vane; LES, Large eddy simulation; PVDF, Polyvinylidene difluoride; RANS, Reynolds-Averaged Navier Stokes; NIOSH, US National Institute for Occupational Safety and Health; WSP, Water-sensitive paper.

E-mail address: davidg@technion.ac.il (D. Greenblatt).

https://doi.org/10.1016/j.expthermflusci.2022.110777  
Received 22 May 2022; Received in revised form 11 August 2022; Accepted 12 September 2022  
Available online 18 September 2022  
0894-1777/© 2022 Elsevier Inc. All rights reserved.
masks are unable to filter aerosols smaller than 5 \( \mu \text{m} \) and are products of all air exhalation such as oral communication, expiration, sneezing and coughing. Most commercial off-the-shelf face masks are approved \([17,18]\) IIR surgical masks. Recent research suggests that controversy concerning rebreathing of CO\(_2\) is associated with breathing \([16]\). There also appears to be some difficulty associated with breathing \([16]\). Many individuals find the wearing of masks to an irritant or even highly uncomfortable. Masks unquestionably inhibit facial recognition and verbal communication, and some people experience “psychological” difficulty associated with breathing \([16]\). There also appears to be some controversy concerning rebreathing of CO\(_2\) in children \([20]\). Transparent face shields were touted as a potential alternative because they produce a physical barrier without covering the mouth and nose. However, virus-laden aerosolized particles can escape below the face shield \([11]\), and their effectiveness depends upon the wearer’s respiration period \([21]\). Furthermore, they are unattractively and tend to become contaminated with exhaled droplets or due to contact with unsterilized surfaces after a short time \([22]\). Recent research has shown, however, that coating the shields with silica nanoparticles can render them more efficacious \([23]\), and modifications to the edges of the shields can improve their effectiveness \([24]\). Despite these essential improvements and studies, there remains an urgent need for a comfortable facemask that does not inhibit breathing and has no negative impact on facial recognition or verbal communication. This article discusses the development and evaluation of such a facemask.

Recently, a fluidic facemask, or ‘Air Curtain’ concept, was proposed by Sakharov and Zhukov \([25]\) involving a rectangular jet of air blown in front of the face. The concept is similar to the well-known and effective concept of an air curtain or “fluidic barrier” at the doorway of a ventilated building \([26]\), and similar concepts are employed in operating theaters \([27]\). A significant advantage is that it serves to protect both wearers from being infected and/or infecting when worn by two individuals. Although the concept has merit, the analysis performed in \([25]\) has several weaknesses. First, the issuing jet is assumed to have a constant width and speed, which is not physically realistic because both laminar and turbulent jets spread to conserve momentum. Second, their analysis does not account for the deflection of the jet due to the relative motion between the jet and the surroundings.

In order to evaluate the concept, a fluidic facemask (or ‘Air-Screen’) was designed and constructed, which embodies a practical and portable implementation of the fluidic barrier by integrating a small fan and power supply into a filter-covered cap. The objective of this study was to experimentally evaluate the Air-Screen concept, based specifically on the practical design described above. Both qualitative and quantitative experiments were performed. Initially, qualitative experiments were used to visualize the main mechanisms of the screen by employing airborne aerosols (\(D \sim 10^3 \mu \text{m}\)), typical of those produced during breathing and verbal communication. Following these encouraging observations, qualitative and quantitative experiments were performed with large droplets (\(D \sim 10^4 \mu \text{m}\)), typical of those produced by sneezing and hard coughing. In all cases, the assumption is that social distancing is not enforced between individuals. Furthermore, a physics-based simplified mathematical model was developed to explain the experimental observations and examine geometric variations.

This paper is structured as follows. The experimental configurations and mathematical model are described in Section 2, the main results are presented in Section 3, model simulation results are presented in Section 4, factors affecting future implementation are presented in Section 5, and conclusions are presented in Section 6.

## 2. Experimental methods and mathematical model

### 2.1. The fluidic facemask

A photograph of the Air-Screen employed in this study is shown in Fig. 1. The outer, visible part of the cap (Fig. 1, left) comprises a thin, flexible frame covered with an antibacterial, porous material to minimize the pressure drop. Two layers of high-efficiency particulate air (HEPA) filter material \([28,29]\) are placed between the material and frame. These can easily be replaced if fouled or replaced with other high-quality standard mask filters \([7]\). The inner part of the cap is made from a comfortable, high-pressure-drop material (Fig. 1, right). The rim is sealed, and there is a gap between the inner and outer parts of the cap that facilitates airflow with minimal pressure losses.

The air screen is produced by a crossflow fan, with an impeller diameter of \(d = 20 \text{ mm}\), mounted in the visor together with an inlet guide vane (IGV) to maximize its efficiency (see Fig. 1, right). According to the manufacturer’s specification, the fan consumes 1.2 W and rotates at \(\omega = 7,400 \text{ rpm}\) to produce a maximum flow rate of \(Q = 5.14 \text{ liters/s}\), corresponding to an exit velocity of 10.2 m/s with a 47.3 Pa suction pressure. In the present setup, pressure losses through the filters reduced the exit velocity to 9.0 m/s. Crossflow fans are ideal for this application because they produce a good balance between pressure and volumetric flow rate, with average specific speeds \([30]\).
crossflow fans can suck air through filters (overcoming the pressure loss) and deliver an effective fluidic barrier jet in front of the face. Moreover, crossflow fans have the ability to produce a rectangular jet and produce low noise due to the structure made of segments of staggered forward-facing blades [30]. Also mounted on the cap visor is a battery charger, an on/off switch and voltage conversion electronics. The total weight of the Air-Screen assembly is Air-Screen 190 g, made up of the cap weights (100 g); and the fan assembly, batteries and filters (90 g).

2.2. Small aerosol setup

Small aerosols were generated from DiEthyl-Hexyl-Sebacate (DEHS) (a colorless and odorless synthetic oil with a density of $\rho_{\text{DEHS}} = 910 \text{ kg/m}^3$) using an ILATEC 40 seeder with six Laskin nozzles. This produced $D = 1.2 \mu \text{m}$ aerosols for a peak Q3 distribution according to the manufacturer’s specification. The seeder was fed into a laminarization chamber via a flexible hose (see Fig. 2, left) consisting of a baffle plate, three turbulence-reducing screens and a bell-mouth contraction leading to a 25 mm nozzle (Fig. 2). The exit bell-mouth contraction velocity in the “top-hat” region was measured using a 1 mm diameter total head probe, and a jet cross flow velocity range $0.8 \text{m/s} \leq U_{\text{jet}} \leq 3.2 \text{m/s}$ was considered. In addition, the volumetric flow was monitored via Dwyer, 0–4 L/minute, rotometer and controlled by both a pressure regulation valve and flow regulators of the seeding generator. The Air-Screen cap was fitted on the mannequin (see Fig. 2), with the fan located 3 cm from the tip of the mannequin nose and at an angle of 20° away from the face, and placed 30 cm from the nozzle outlet.

Two methods were employed for flow visualization using high-speed and standard video recordings. For the first method, high-speed visualization was performed at 300 frames per second using a Phantom v7.3 camera mounted on the right side of the mannequin. A 180 W UV diode cannon, mounted below the mannequin, was used to illuminate the aerosols in the vicinity of the mannequin. A black synthetic mat was mounted on the mannequin’s left side to minimize reflections and create a sharp contrast between aerosols and the surrounding air (see Fig. 2). For the second method, standard recordings at 30 frames per second were performed by illuminating the vertical plane below the mannequin using a light sheet generated by a 100-milliwatt laser (532 nm). The light sheet was aligned with the center plane of the mannequin and was used to produce video recordings that are included as supplemental material. Experiments performed without the Air-Screen active are referred to as “baseline” conditions.

2.3. Large droplet setup

Large droplets were generated utilizing a Paasche VL1007 hollow cone nozzle airbrush with a needle of size 0.7 mm. The high-pressure inlet to the nozzle was controlled via a pressure regulating valve and produced an exit velocity of up to 50 m/s, measured at 1 cm from the nozzle exit. A syringe, with the plunger removed, was used to feed the airbrush, and the water flowrate was monitored at approximately 0.005 L/minute. The airbrush was set at an angle of 10° to the horizontal and located at 100 cm from the mannequin. A second Air-Screen fan was mounted 4 cm and 10 cm in front of and above the airbrush, respectively.

In this experimental setup, the airbrush was considered the...
Experimental Thermal and Fluid Science 141 (2023) 110777

4

were acquired using two independent methods, one qualitative and one using a WPS spreading to droplet size factor [31]. In so doing, the WPS size/pixel ratio. The actual droplet size was then determined. Due to the WSP and scanner limitation, the minimum detectable droplet size was approximately $D = 60 \pm 10 \mu m$, determining the lower limit of identifiable droplets. By using several masking and threshold parameters, errors of $\varepsilon < 5\%$ were realized, both in frequency and size for droplets larger than 100 $\mu m$. The estimated error is higher for smaller droplets since pixelization impact increases for smaller contours due to reduced edge segmentation precision. The result presented in 3.2 are considered conservative, since the percentage of blocked droplets increases as the droplet size decreases.

2.4. Mathematical model

In an attempt to explain the experimental observations and suggest improvements, a physics-based mathematical model was developed to predict the aerosol and droplet trajectories when encountering the Air-Screen jet. The limitations in the physical formulation by Sakharov & Zhukov (2020), were overcome by employing a turbulent jet [32], modified by a weak crossflow [33]. Furthermore, the spectrum of virus-laden aerosols and droplets, corresponding to the experiments was assumed. Sections 2.4.1–2.4.2 describe the theory of two-dimensional (2-D) turbulent jets in a crossflow and the forces acting on droplets and aerosols, and Section 4 presents the results of the numerical simulations.

2.4.1. Turbulent Jet-in-a-Crossflow

The theoretical development of 2-D turbulent jets relies on conservation of momentum [32], together with the solution of the boundary layer equations, resulting in the velocity component in the streaming direction ($x$):

$$u = u_c sech^2(\eta)$$

and the crossflow direction ($y$):

$$v = u_c (2\eta sech^2 \eta - tanh \eta)/2$$

where $\eta = \sigma (y/x)$, $u_c \sim (x - x_0)^{-1/2}$ is the centerline velocity, and $\sigma$ and $x_0$ are empirical constants (see definitions in Fig. 5). The interaction of the fluidic barrier with virus-laden aerosols and droplets can be modeled as: (a) the interaction between rigid bodies (droplets) and a fluidic jet; (b) the interaction between a fluidic jet and a virus-laden crossflow; or (c) a combination of the two. Drikakis (2020a) show that droplets and aerosols produced during a strong cough cannot travel more than 2 m in distance, where the vast majority will not be carried more than 1 m, and are primarily spread via airborne transport. This correlates to the study of Wang et al. (2021), estimating that only a small percentage of infection is carried via large droplet infection. Thus, a more accurate method to model virus-laden and Air-Screen interaction is between a fluidic jet and an airborne virus-laden crossflow.

Three-dimensional and two-dimensional turbulent jets in crossflows were studied extensively [33–38]. Huang et al. (2005) classify 2-D jets in a crossflow into two main categories, namely 2-D discharges in weak or strong crossflow. Irrespective of the crossflow strength, the jet is deflected due to two main mechanisms, namely a momentum balance between the jets and a pressure gradient across the jet that is expressed as a “drag” force. Thus, the trajectory of the jet’s centerline can be expressed as:

$$\frac{dy}{dx} = \frac{U_e (Q - Q_e) + (C_d/2)U_e^2 \delta}{U_e h}$$

where $U_e$ is the crossflow velocity, $(Q - Q_e)$ is the difference between slot and local downstream volume flux, $C_d$ is the drag coefficient, and $h$ is the slot width. However, a turbulent jet discharging into a very weak crossflow resembles a turbulent jet discharging into an ambient fluid (see Fig. 5b). For these weakly advected jets, it can be assumed that the only interaction between the flows is a momentum transfer which is valid under the assumption [33]:

$$\frac{x}{U_e h/\delta} \leq 0.4$$

and for these flows, we can assume that $C_d \approx 0.8$ [33].

The boundary conditions employed in the model corresponded directly on the Air-Screen (see Section 2.1), namely, $U_0 = 9 \text{ m/s}$ and $h = 5 \text{ mm}$. The high fan aspect ratio (20) justified our two-dimensionality assumption. Several velocity profiles downstream of the jet exit were
Fig. 4. Example of a 150 mm $\times$ 120 mm piece of WSP on the right and the results of the image segmenter used to produce an image mask of droplet size impacting the WSP on the left. Magnified rectangular sections are shown on the right.

Fig. 5. (a) Schematic of a two-dimensional turbulent jet; (b) schematic of a 2-D weakly advected turbulent jet in a crossflow.

Fig. 6. Non-dimensionalized turbulent velocity profiles with associated error bars, measured at 3 cm and 7 cm downstream of the fan exit plane, shown together with the theoretical result shown in equation (1).
surveyed with a small flattened pitot tube in conjunction with a Lambrecht micro-manometer with a ±700 Pa pressure range and a ±0.2 Pa resolution. An example of two velocity profiles, together with the theoretical results (see equation (1)), are shown in Fig. 6 and indicate a close correspondence based on the jet spreading rate constant $\sigma = 7.67$. The error bars shown in the figure are based on the micro-manometer resolution error.

2.4.2. Forces acting on aerosols and droplets

The range of virus-laden aerosol and droplet diameters varies between $\alpha(10^5)$ to $\alpha(10^7)$ μm, and this results in a relatively large Reynolds numbers range ($Re \approx 10^5$) where:

$$Re \equiv \frac{|\vec{U}_{vel\text{-drop}}| D}{v_a}$$

(5)

$|\vec{U}_{vel\text{-drop}}|$ is the relative velocity magnitude between the aerosol velocity vector $\vec{U}_{drop}$ and the surrounding flow $\vec{U}$ and $v_a$ is the air kinematic viscosity. This results in more than an order of magnitude variation in the aerosol or droplet’s drag coefficient $C_D$ [39]. To account for the large Reynolds number variation, the approximation attributed to Brown et al. [40] was used, namely:

$$C_D = \frac{24}{Re} \left(1 + 0.15(Re^{0.88})\right) \left(1 + \frac{0.407}{1 + 8710/Re}\right)$$

(6)

which is valid for $Re \approx 10^4$. In order to compute the drag force, the relation:

$$|\vec{F}_D| = \frac{\pi}{8} C_D \rho_a |\vec{U}_{vel\text{-drop}}|^2$$

(7)

was used, where $\rho_a$ is the air density. In addition, forces due to gravity and buoyancy were calculated according to:

$$\vec{F}_b = \frac{\pi}{6}(\rho_a - \rho_i) \vec{U} D^3 \approx \frac{\pi}{6} \rho_i R D^3 \vec{\hat{x}}$$

(8)

where $\rho_i$ is the density of the aerosols and droplets [41].

3. Result and discussion

3.1. Small aerosols

Initial experiments were designed to visualize the ability of the Air-Screen to shield against small size aerosols, $D \sim 10^6$ μm, as described in Section 2.2. Fig. 7 shows pairs of raw image pairs arbitrarily extracted from the high-speed camera recordings for baseline and activated Air-Screen cases (left and right). Four cases are shown where the jet velocity increased from 0.8 m/s to 3.2 m/s. It is important to note that these relatively high jet velocities exceed those that transpire during ordinary oral communication at these distances [42]. These experiments also represent an individual working, walking or jogging in a field of aerosol-laden air. This can occur in environments containing small aerosols (less than 5 μm), where pathogens predominate and can remain airborne in still air for long periods without sinking to the ground or evaporating [14].

In all baseline cases, the face of the mannequin is effectively engulfed by the aerosol, where greater aerosol concentrations are present with increasing jet velocity. This is simply a consequence of more aerosol-laden air diluting with the ambient air. In contrast, a clear demarcation can be seen between the clean air and the aerosol-laden air when the Air-Screen is activated. This is an unambiguous effect of the Air-Screen, as shown in Fig. 8, where all images are averaged. These images show a clear contrast between the baseline and activated cases. The averaged images were processed based on the brightness distribution of the captured data, using a segmentation code and based on the camera’s optical sensor to evaluate the borders satisfying 97% ± 2% of the aerosol density. The low precision of the aerosol density borders is due to the changing brightness field each

![Fig. 7](image-url)
particle emits when exposed to a monochromatic light source. Nevertheless, this technique facilitates a precise method to qualify the effectiveness of the Air-Screen against small aerosols.

The choice of DEHS for flow visualization was based on its long droplet life and its optical illumination properties under UV light. Although saliva contains approximately 99% water [43], and is therefore used in the droplet experiments (see Section 3.2), it was determined that particle drag based on its diameter and associated Reynolds number were of paramount importance and that viscosity and surface tension were secondary parameters. In future research, aerosols should be generated by a three-jet collision nebulizer and fed into a Goldberg drum similar to that used in [44].

3.2. Large droplets

Following the small aerosol experiments, extensive qualitative and quantitative droplet transmission experiments were conducted. For the qualitative experiment, the droplets were recorded at 1000 frames/second, five times, over a one-second interval at a time. Using a dedicated image processing program, the pathlines of the droplets were processed and the results are shown in Fig. 9. Note that only droplets that pass through the 1 mm thick laser sheet are illuminated and recorded. The figure is arranged to show the baseline case (all Air-Screens off) and the other three combinations. It can be seen that when the mannequins’ Air-Screen is activated, the number of water particles passing through the sheet diminishes. However, when the transmitter Air-Screen is activated, the number of particles is diminished virtually to zero. This is a clear demonstration that the Screen can protect others from an infected wearer. With both Screens active, the situation remains visually the same.

The observations made in the qualitative experiments are reflected in the quantitative WPS results. Fig. 10 shows an example of a single set of...
The WPS experiments, out of 24 different experiments made with various droplet sizes. The black dots on the yellow WPS indicate where the water droplets have impacted the paper. Note that the actual aerosol size is smaller than the observed dot due to the spreading factor [31] explained in Section 2.3. It should also be appreciated that the WPS is placed relatively low, corresponding to the position of the nose and mouth, where the Air-Screen jet velocity is relatively low. Despite this, the Air-Screen is remarkably effective when worn by the transmitter.

This data was quantified using the image segmentation program, described in Section 2.3, to extract the droplet’s size and apply the spreading corrections. The results are presented in the form of a histogram in Fig. 11, and can be classified into two groups: (i) experiments with a droplet size of $D > 150 \mu m$, and (ii) experiments with a droplet size of $D < 150 \mu m$. This demarcation reveals several important aspects. When the mannequin Air-Screen is activated, it eliminates more than 62% of all the droplets corresponding to $D < 150 \mu m$. This shows the efficacy of the Air-Screen not only for shielding against aerosols but also as a partial shield against relatively large droplets. As anticipated, when the transmitter’s Air-Screen is activated, it blocks more than 99% of high momentum droplets issued at a distance of 100 cm. When both Air-Screens are active, 99.8% of the droplets are eliminated (see the inset in Fig. 11). In fact, only a single droplet, of diameter $D = 330 \mu m$, impacted the WSP out of thousands that impacted without the Air-Screen active. Hence, the key to success against even the largest droplets appears to be the Air-Screen worn by a transmitter. The transmitter Screen diverts the larger droplets sufficiently downwards such that they are far less likely to impact an individual’s face at a similar height.

To further understand the implication of the experimental data, a visually intuitive representation is shown in Fig. 12. The facial image

![Fig. 10. Scanned images of unprocessed WSP for a single set of experiments. The black dots reflect the impact of the water droplets on the paper.](image1)

![Fig. 11. Histogram of the WSP data that was processed using an image mask to extract droplets’ size and the corrections for spreading. (right) medium-large droplet experimental setup, $\bar{D} = 93 \mu m \pm 24 \mu m$. (left) very large droplet experimental setup, $\bar{D} = 273 \mu m \pm 131 \mu m$. (Inset: magnified area showing effects of transmitter and transmitter-mannequin Air-Screens active).](image2)
shown in the figure is based on a synthetic image generated by StyleGAN code [45] and corresponds to an archetypal face for illustrative purposes. The image represents the location of the WPS, illustrating a slightly larger morphological facial extent than average [46]. The figure exemplifies the distribution of the droplet penetration of the facial area with respect to the downstream direction as gained from the experiments with a droplet size of up to 200 μm. It is evident that for large droplets, the target Air-Screen deflects the droplets downward, thereby providing greater protection of the eyes and nostrils, due to the higher jet velocity closer to the fan. As can be inferred from the above results shown in Fig. 9 to Fig. 11, these images graphically illustrate the protection that the Air-Screen affords the target, particularly when worn by a transmitter. In conclusion, the Air-Screen provides excellent protection against high momentum droplets to the individual when worn by a transmitter and high protection when only used by the target.

Although the blocking effect is clearly demonstrated, further experiments are crucial to properly evaluate the Air-Screen. Ideally, if all individuals wear the Air-Screen, e.g. in an office or factory setting, then every-one is equally protected. If a single individual wears the Air-Screen, then it must be clearly shown that non-wearers are not harmed. In both instances, the Air-Screen must be evaluated against other gold-standard mask efficacy measurements to enable a comparison with standardized assessment tools. Typically, mask effectiveness is measured using the difference of particle count and sizing down to 300 nm. Such an assessment must be done for both outward and inward penetration or dissemination and can be thus assessed for both exhalation and inhalation configurations, i.e., protection of the transmitter and wearer. Finally, the comfort and practicality of the Air-Screen were considered in a pilot study conducted by ten human subjects. The main results of the study, as well as suggested improvements, are discussed in Section 5.

4. Numerical simulation and results

At the outset, the combined velocity field of the jet and crosswind was solved numerically across the entire domain, as explained in Section 2.4.1. Droplet size, location and velocity boundary conditions (Cauchy boundary conditions) were selected. Thus, the initial conditions \( t = 0 \) for the simulations were the weakly advected jet velocity field together with the specific droplet parameters and boundary conditions. An explicit fourth-order Runge-Kutta method [47] was used to approximate the aerosol or particle locations at each time step, \( \Delta t \), based on the current values and a weighted average of four increments. For each increment, based on the droplet’s location and its prior velocity vector, the relative velocity vector was determined, thereby facilitating the calculation of \( Re \) and hence \( C_D \). On the basis of these results, the droplet acceleration vector was estimated, and its location at \( t + \Delta t \) was calculated. The accumulated error of the model is on the order of \( \Delta t^4 \); thus, the implementation of small \( \Delta t \) ensured that the numerical errors were secondary compared to the model assumption errors.

Although the geometric parameters, \( y_{\text{jet}} \) and \( \Delta \alpha_{\text{jet}} \) (see Fig. 13, left) were fixed at 3 cm and 20° in the experiments, these parameters were varied in the mathematical model, bracketing the parameter space defined by \( 1 \mu m < D < 250 \mu m \), 1 cm \( < y_{\text{jet}} < 5 \) cm and 10° \( < \Delta \alpha_{\text{jet}} < 30° \) (see Fig. 13, right). The plot identifies the maximum airborne droplet diameter effectively blocked by the jet as a function of \( y_{\text{jet}} \) and \( \Delta \alpha_{\text{jet}} \), where body forces are neglected (\( \vec{F}_b = 0 \)). Neglecting body forces produces conservative results because when they are included, gravity dominates over buoyancy and their net effect is downwards. Droplets are considered effectively “blocked” when their penetration of the jet falls below 15 cm from the jet slot height. The distance of 15 cm is based on the average of male and female anthropometric measurements between the mid-forehead and base of the chin, encompassing many different countries and ethnic groups [46]. The bright area in the lower left-hand part of the contour plot indicates poor effectiveness against airborne aerosols and droplets, which penetrate across the jet. Relatively large increases in either of these parameters, e.g. \( y_{\text{jet}} = 3 \) cm or \( \Delta \alpha_{\text{jet}} = 20° \), are only able to block the smallest aerosols. However, their combined effect under identical conditions effectively blocks droplets up to 100 μm, which is considered the upper minimum diameter of virus-laden airborne aerosols [19]. Further simultaneous increases in these parameters further increase the diameter of droplets blocked.

Although increases in the distance \( y_{\text{jet}} \) effectively block large aerosols and particles, there is a practically viable limit for which the fan can be comfortably and ergonomically incorporated into a cap. In addition, if \( y_{\text{jet}} \) is too large, then the face is exposed to the possibility of virus-laden aerosols in yawed crosswinds that can obliquely enter the region between the fan and face. Regarding increases in \( \Delta \alpha_{\text{jet}} \), there is also a practical limit determined by high levels of turbulent diffusion in the lower part of the jet which increases the possibility of virus-laden aerosol penetration in this region. Accordingly, we can define an “optimum design space” (shown in Fig. 13) that maximizes the efficiency of the Air-Screen against large aerosols of \( D \approx 100 \mu m \) [19]. It can be seen that the “operating point” used in our experiments, falls comfortably within the optimum design space.

Visualizations of droplets pathlines embedded in a crossflow velocity field of \( U_b = 1.5 \) m/s, where the acquired jet velocity field corresponding to \( y_{\text{jet}} = 3 \) cm and \( \Delta \alpha_{\text{jet}} = 20° \), is shown in Fig. 14. The boundary condition \( \bar{u}_{\text{inflow}} = \bar{U}_b \) along the left inflow boundary at 20 cm from the model was employed. The trajectories of three different droplet sizes are shown, namely \( D = 5 \) μm, 100 μm and 300 μm, both neglecting and including...
body forces. It can be seen that with no fan active (Fig. 14a, \( \overline{F}_{b} = 0 \)), all particles impact the face. Small particles of 5 \( \mu m \) (Fig. 14b, \( \overline{F}_{b} = 0 \)) are deflected to a relatively large distance from the face because, due to their low mass, they are easily accelerated (curved away). Further increases in diameter to 100 \( \mu m \) (Fig. 14c, \( \overline{F}_{b} = 0 \)) show reduced effectiveness as the acceleration is diminished, but no penetration is observed. For 300 \( \mu m \) droplets (Fig. 14d, \( \overline{F}_{b} = 0 \)), even though droplets penetrate the jet, those approximately in line with the mouth can be seen to be progressively deflected to a relatively large distance from the face because, due to their low mass, they are easily accelerated (curved away). For 300 \( \mu m \) droplets (Fig. 14e, \( \overline{F}_{b} = 0 \)), even though droplets penetrate the jet, those approximately in line with the mouth can be seen to be progressively deflected to a relatively large distance from the face because, due to their low mass, they are easily accelerated (curved away). For 300 \( \mu m \) droplets (Fig. 14f, \( \overline{F}_{b} = 0 \)), even though droplets penetrate the jet, those approximately in line with the mouth can be seen to be progressively deflected to a relatively large distance from the face because, due to their low mass, they are easily accelerated (curved away).

For our aerosol experiments (Section 3.1), we considered \( D = 1.2 \mu m \) under the conditions described above, where crossflow speeds \( U_{b} \) of 0.8 m/s, 1.5 m/s, 2.4 m/s and 3.2 m/s were considered. This range was designed to cover speeds of fast walking or typical movements in an indoor environment to fast jogging. Aerosol pathline results of numerical simulations corresponding to these conditions are shown in Fig. 15. These results indicate that blockage is effective up to \( U_{b} = 2.4 \) m/s, a typical jogging speed (8.6 km/hour). There is a clearly a qualitative similarity between these results and the experimental visualizations shown in Fig. 8.

The results presented above correspond to a single fan operating point where the fan momentum flux is \( J_{0} = 31 \) mN. Based on our measurements presented in Section 2.4.1. However, the fan speed (input power) can be controlled, and as a result, the momentum flux or jet velocity distribution can be varied to accommodate different environmental conditions. Fig. 16 illustrates the effect of the momentum flux on the minimum airborne droplet diameter necessary to penetrate through the fluidic barrier compared to the nominal operation regime for the parameters used in Fig. 14 and Fig. 15. There is an almost linear relationship between the efficacy of the fluidic barrier and the momentum flux, and clearly, this motivates in favor of greater power input. However, this will impact noise and power consumption (battery capacity) which is proportional to \( J_{0}^{2} \). Future research should aim to increase the system’s efficiency by adapting the fan momentum flux, and hence input power, to the prevailing surrounding conditions.

The numerical simulation results presented above corresponded qualitatively with the experiments in a time-averaged sense. Note, however, that the model is two-dimensional in that it does not account for three-dimensional (jet-edge) effects or flows approaching the face at an angle (yawed flows). The geometric (face) boundary condition is also not explicitly modeled, nor are the effects of breathing. Moreover, the model assumes that the water droplets are solid spheres and therefore does not account for evaporation at the air–water interface [12]. These limitations should be addressed by computational studies employing either Reynolds-Averaged Navier Stokes (RANS) equations or Large Eddy Simulations (LES).

5. Factors affecting implementation

The present investigation was the first ever conducted on a wearable fluidic protection device of this type, and therefore experiments and modeling were, by design, preliminary. The overarching conclusion was that the concept has merit. Nevertheless, both scientific and practical factors must be addressed before the Air-Screen can be considered a viable alternative to facemasks. These factors are discussed below.

5.1. Scientific factors

From the scientific perspective, although our experiments determined that the Air-Screen is potentially feasible, future gold-standard measurements must be performed that measure incoming particle size distribution and quantity from the transmitter versus that at the inlet of the mannequin. To achieve this, mannequins that simulate human breathing must be employed, both as the transmitter and the wearer. Another limitation of this research was that experiments were only performed with the mannequin or target perpendicular to the incoming flow or transmitter. Clearly, this is a special case, and further experiments must be performed to quantify the effect of the incoming flow relative to the mannequin (yaw angle).

The present flow visualizations produced by illuminating the seeded air with UV light produced an aggregate or average of the particle’s interaction with the mannequin. These data must be complemented with higher quality fluid visualization [11,48] and flowfield measurements [49,50]. The diffusion and evaporation of the aerosols and droplets by the Air-Screen must also be investigated. At present, our hypothesis is that the Air-Screen increases the diffusion of already existing aerosols and droplets and also enhances evaporation, but this must be verified experimentally. Finally, the results of these investigations must be compared with those of standard protective equipment, particularly N95 facemasks or standard approved IIR surgical masks [17,18].

Investigating the impact of various filters/cap covering materials is essential. On the one hand, HEPA/ULPA filters can filter the majority of bacteria and viruses carried by aerosols in the air [51]. On the other hand, the outer porous cover of the cap was selected to minimize the
pressure drop, thereby avoiding the need for a more powerful fan and preventing the filter from being damaged. At present, the Air-Screen does not integrate a virucidal system because this investigation was limited to evaluating the Air-Screen’s ability to protect individuals from one another without enforcing social distancing. This can be rectified using novel virucidal materials for filtration, such as charged PVDF \cite{52} or HEPA-ClO$_2$ filters \cite{53}. Hydrophilic porous materials should also be examined, as they can decrease the survivability of viruses through capillary imbibition \cite{54}, while maintaining a minimal pressure drop. There may also be an advantage to adding a second porous material downstream of a HEPA filter \cite{55}.

5.2. Practical factors

In terms of practical application, a pilot study was conducted that employed ten human subjects who evaluated the Air-Screen with respect to comfort, weight, convenience and noise. The feedback was mainly positive concerning comfort and weight, primarily because the entire weight of the Air-Screen (cap, fan assembly, batteries and filter) is 190 g and, in terms of comfort, has the same sensation as a baseball cap. The subjects did not experience undue vibrations because the already small vibrations were absorbed by the fabric covering the visor. The subjects also did not experience any eye discomfort while remaining stationary or walking. This is because the jet width is thin (5 mm), it is spaced 3 cm from the forehead, and it is deflected at $20^\circ$ away from the face (see Section 4).

Some of the subjects found that the fan noise was distracting or annoying. Fan noise is generally broadband and results from turbulence and vortex generation, together with superimposed pure tone components related to the geometry and fan-blade speed. In future research, we shall attempt to reduce the fan’s noise by increasing the impeller diameter and reducing the rotation speed while maintaining constant momentum flux. This strategy, which we have begun implementing, is described below.

Fig. 14. Numerical simulation of particle trajectories at a crossflow velocity of $U_a = 1.5 \text{m/s}$ and fan parameters: $U_0 = 9 \text{m/s}, d = 5 \text{mm}, y_{jet} = 3 \text{cm}$ and $\alpha_{jet} = 20^\circ$ (a–d: gravitational forces neglected in simulations; e,f: gravitational forces included in simulations).
On the basis of an extension of Lighthill’s general theory of aero-
dynamic sound and dimensional analysis [56], it can be shown that the
change in the sound pressure level depends on the ratio of the impeller
tip speeds, as follows:

\[ \Delta \text{SPL} = 60 \log \left( \frac{\omega_{\text{new}} d_{\text{new}}}{\omega d} \right) \] (9)

In order to maintain the same operating point volumetric flowrate
and slow the rotational speed, we require the fan performance map.
However, no fan performance map was available, therefore we con-
structed two standalone calibration facilities for both the fan calibration
and the cover-plus-HEPA filter covering (system load). Our fan cali-
bration data (filled symbols) together with the manufacturer’s speci-
cation data (open symbols, see Section 2.1) are shown in Fig. 17,
together with the system load data. The curve through the fan data
points is a second-order least-squares curve.

In order to make estimates of the rpm reduction while still remaining
at or above the operation point, we propose to increase the impeller
diameter \( d \) while maintaining geometric similarity. For the evaluation,
well-known fan laws for flowrate were used for pressure and power [57],
namely:

\[ \phi = \frac{Q}{\omega d^3} \] (10)
\[ \psi = \frac{p}{\rho (\omega d)^2} \] (11)
\[ \Pi = \frac{W}{\rho \omega^3 d^6} \] (12)

together with the least-squares curve, to make estimates of the new

Fig. 15. Numerical simulation of aerosol \((D = 1.2 \mu m)\) trajectories of at various crossflow velocities, where \(U_0 = 9 \text{ m/s}, h = 5 \text{ mm}, y_{\text{jet}} = 3 \text{ cm} \) and \(\alpha_{\text{jet}} = 20^\circ\). Body forces are included in the simulations but have a negligible effect on \(D = 1.2 \mu m\) aerosols.

Fig. 16. Numerical simulations results showing the effect of momentum flux on the minimal airborne droplet diameter necessary to penetrate through the fluidic barrier, referenced to \(D_0 = 100 \mu m\) and \(J_0 = 31 \text{ mN}\). Fan and crosswind parameters \(y_{\text{jet}} = 5 \text{ cm}, \alpha_{\text{jet}} = 20^\circ\) and \(U_{\text{a}} = 1.5 \text{ m/s}\).
Fig. 17. Original fan performance curve and system curve, together with fan performance estimates for a 20% larger geometrically similar impeller, based on fan laws described in equations (10)–(12).

operating conditions. For illustration purposes, we make predictions based on increasing the impeller diameter by 20% and decreasing the rotational speed by 20%, while maintaining the same tip speed as shown in Fig. 17. This results in a higher volumetric flowrate for a given pressure but at the cost of 44% more input power. This may also result in noise reduction because, even though the blade tip speed is the same, the rotational speed is lower.

The next step of the strategy is to further reduce the rotational speed of the new impeller until both an acceptable noise level is reached and the original jet momentum flux, proportional to \( d\omega_0^2 \), is attained. In the example shown in the figure, the rotational speed is reduced by a factor of 1.31 to attain the same momentum flux. By substituting these new values for \( \omega_{\text{new}} \) and \( d_{\text{new}} \) into equation (9), we can potentially attain a 2.4 dB noise reduction with 10% additional power. Note that this new impeller and rotational speed produces a lower shutoff pressure but a higher operating point pressure. The lower shutoff pressure is unimportant because the fan is operated only near its maximum flowrate.

6. Conclusions & recommendations

This paper presented the first experimental evaluation of a fluidic facemask, or Air-Screen. The design philosophy was described, and experiments were performed to evaluate its potential for protection against virus-laden aerosol and droplet infections, such as SARS-CoV-2 (COVID-19). The experiments included both small-sized airborne aerosols \((D \sim 10^3 \mu m)\) and large droplets \((D \sim 10^2 \mu m)\) generated by sneezing or hard coughing from a 1-meter distance.

For small aerosols, without the Air-Screen active, the face of the mannequin is entirely engulfed in airborne particles that can impact the eyes and even wearing a surgical mask does not eliminate the danger of inhaling the virus-laden aerosols. In contrast, the Air-Screen produced a clear demarcation between clean aerosol-free air and aerosol-laden air at ambient conditions and for wind speeds of up to 2.4 m/s (fast walking speed). For large droplets, the Air-Screen eliminated most of the droplets with a diameter of less than 150 \( \mu m \) produced by a “transmitter”. With the transmitter Air-Screen activated, 99% of all droplets were eliminated, and with both Air-Screens active, virtually all droplets were eliminated. Based on a simplified physics-based mathematical model, numerical simulations were employed to gain greater insight into the experimental results and as a development tool for future studies. The model results illustrated that the selected experimental geometry fell within the optimum design space for blocking aerosols and droplets.

The main conclusion of this evaluation is that the Air-Screen is remarkably effective in blocking aerosols and droplets from the face of the wearer. Nevertheless, future studies must quantify the Air-Screen’s effectiveness using gold-standard test methods. These should include a “breathing mannequin” where the inhaled number and aerosol size distribution are directly measured. Three-dimensional effects should be introduced by directing aerosols and droplets at oblique angles to the mannequin. This will facilitate a comparison between the Air-Screen and other commercial facemasks. In parallel, high fidelity CFD models, such as large eddy simulations (LES), should be employed to simulate the full three-dimensional problem and provide a solid foundation upon which to make design modifications and innovations. Feedback from a pilot study showed that Air-Screen is comfortable due to its low mass (less than 200 g), but that efforts should be made to reduce fan noise.

Competing interests. None.

This research received no specific grant from any funding agency, commercial or not-for-profit sectors.

Ethical standards. The research meets all ethical guidelines, including adherence to the legal requirements of the study country.

Credit authorship contribution statement

David Keisar: Methodology, Data curation, Visualization, Software, Writing – review & editing. Anan Garzozi: Methodology, Data curation, Visualization. Moshe Shoham: Conceptualization, Funding acquisition.

David Greenblatt: Methodology, Data curation, 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.
Data availability
Data will be made available on request.

Appendix A. Supplementary data
Supplementary data to this article can be found online at https://doi.org/10.1016/j.expthermfusci.2022.110777.

References
[1] CDC. Science Brief: Community Use of Cloth Masks to Control the Spread of SARS-CoV-2 (2021). https://www.cdc.gov/coronavirus/2019-ncov/science/briefs/masking-sars-cov2.html (accessed June 11, 2021).
[2] L. Bourouiba, E. Dehandschoewcker, J.W.M. Bush, Violent respiratory events: On coughing and sneezing, J. Fluid Mech. 745 (2014) 537–563, https://doi.org/10.1017/jfm.2014.88.
[3] D.K. Chu, E.A. Eld, S. Duda, K. Solo, S. Yasasch, H.J. Schünemann, A. El-harakeh, M.J. Katic, C.R. Forrest, K.W. Alt, A. Malinowski, M. Negasheva, S. Manolis, M.G. Gioia, H. Zhao, I. Neumann, J. Chan, J. Smith, E. de Wit, V.J. Munster, Aerosol and Surface Stability of SARS-CoV-2 as a Function of Relative Humidity and Temperature, PLoS Pathog. 16 (6) (2020) e1009178, https://doi.org/10.1371/journal.ppat.1009178.
[4] K.P. Fennelly, Particle sizes of infectious aerosols: implications for infection control, Lancet Resp. Med. 8 (2020) 914–924, https://doi.org/10.1016/S2213-9249–2020-00813-3.

https://doi.org/10.1016/j.expthermfusci.2022.110777.
[50] A. Agrawal, R. Bhardwaj, Reducing chances of COVID-19 infection by a cough cloud in a closed space, Phys. Fluids. 32 (2020), 101704, https://doi.org/10.1063/5.0029186.

[51] D.T. Liu, K.M. Phillips, M.M. Speth, G. Besier, C.A. Mueller, A.R. Sedaghat, Portable HEPA Purifiers to Eliminate Airborne SARS-CoV-2: A Systematic Review, Otolaryngol. - Head Neck Surg. (United States) 166 (2022) 615–622, https://doi.org/10.1177/01945998211022636.

[52] W.W.F. Leung, Q. Sun, Charged PVDF multilayer nanofiber filter in filtering simulated airborne novel coronavirus (COVID-19) using ambient nano-aerosols, Sep. Purif. Technol. 245 (2020) 1–12, https://doi.org/10.1016/j.seppur.2020.116887.

[53] C. Yan Suen, Y. Tak Lai, K. Hei Lui, Y. Li, H. Hoi Ling Kwok, Q. Chang, J. Hong Lee, W. Han, X. Yang, Z. Yang, Z. Mo, P. Kin Shing Wong, A. Chi Tat Leung, J. Kai Cho Kwan, K. Lun Yeung, Virucidal, bactericidal, and sporicidal multilevel antimicrobial HEPA-CIO2 filter for air disinfection in a palliative care facility, Chem. Eng. J. 433 (2022) 134115. https://doi.org/10.1016/j.cej.2021.134115.

[54] S. Chatterjee, J.S. Muralidharan, A. Agrawal, R. Bhardwaj, Why coronavirus survives longer on impermeable than porous surfaces, Phys. Fluids. 33 (2021), 021701, https://doi.org/10.1063/5.0037924.

[55] Y. Narayan, S. Chatterjee, A. Agrawal, R. Bhardwaj, Assessing effectiveness and comfortability of a two-layer cloth mask with a high-efficiency particulate air (HEPA) insert to mitigate COVID-19 transmission, Phys. Fluids. 34 (2022), 061703, https://doi.org/10.1063/5.0094116.

[56] N. Curle, The influence of solid boundaries upon aerodynamic sound, Proc. R. Soc. London. Ser. A. Math. Phys. Sci. 231 (1955) 505–514, https://doi.org/10.1098/rspa.1955.0191.

[57] B.R. Munson, D.F. Young, T.H. Okishi, Fundamentals of fluid mechanics, 7th ed., Wiley, 2012. https://doi.org/10.1201/b11709-7.