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.
Computational design and experimental analysis of a novel visor for COVID-19 patients receiving high-flow nasal oxygen therapy

Masooma Ijaz a, Sorcha Ni Fhrighil a, Rory Brett a, Jack Connolly b, Alan Conneely c, Gerard O’Connor d, Martin O’Halloran d, Sajjad Yousefian a,∗

a Mechanical Engineering, College of Engineering and Informatics, National University of Ireland, Galway, Ireland
b Bioinnovate Ireland, National University of Ireland, Galway, Ireland
c National Centre for Laser Applications, School of Physics, College of Science and Engineering National University of Ireland, Galway, Ireland
d Translational Medical Device Lab, Lambe Institute for Translational Research & HRB Clinical Research Facility, University Hospital Galway, Galway, Ireland

A R T I C L E   I N F O

Article history:
Received 18 January 2022
Received in revised form 16 September 2022
Accepted 24 September 2022
Available online 30 September 2022

Keywords:
COVID-19
High flow nasal oxygen therapy (HFNOT)
Computational fluid dynamics (CFD)
Particle extraction visor
Schlieren imaging

A B S T R A C T

The Covid-19 global pandemic has reshaped the requirements of healthcare sectors worldwide. Following the exposure risks associated with Covid-19, this paper aims to design, optimise, and validate a wearable medical device that reduces the risk of transmission of contagious droplets from infected patients in a hospital setting. This study specifically focuses on those receiving high-flow nasal oxygen therapy. The design process consisted of optimising the geometry of the visor to ensure that the maximum possible percentage of harmful droplets exhaled by the patient can be successfully captured by a vacuum tube attached to the visor. This has been completed by deriving a number of concept designs and assessing their effectiveness, based on numerical analysis, computational fluid dynamics (CFD) simulations and experimental testing. The CFD results are validated using various experimental methods such as Schlieren imaging, particle measurement testing and laser sheet visualisation. Droplet capturing efficiency of the visor was measured through CFD and validated through experimental particle measurement testing. The results presented a 5% deviation between CFD and experimental results. Also, the modifications based on the validated CFD results improved the visor effectiveness by 47% and 38% for breathing and coughing events, respectively

© 2022 The Author(s). Published by Elsevier Masson SAS. This is an open access article under the CC BY license (http://creativecommons.org/licenses/by/4.0/).

1. Introduction

The Covid-19 disease is caused by the Severe Acute Respiratory Syndrome Coronavirus 2 (SARS-CoV-2) virus [1]. This is a highly contagious disease, susceptible to person-to-person transmission by airborne droplets and aerosols. Since the onset of the virus, the World Health Organisation (WHO) has declared the Covid-19 a worldwide pandemic, in part due to its high rate of transmission [2].

Covid-19 transfers from an infected individual’s mouth and nose through aerosols and droplets dispersed during breathing, coughing, sneezing and speaking. Numerous studies relating to understanding the etymology and spread of the virus, have been carried out, including studies on the efficacy of different preventive measures taken to reduce the spread of the virus [3]. This includes wearing of Personal Protective Equipment (PPE) by the healthcare staff and face masks by the general public. Despite the efforts of medical staff worldwide to follow recommended procedures to minimise infection spread in hospital settings, the spread of the Covid-19 virus was rampant in many healthcare environments worldwide throughout 2020 [4].

To date, mechanical ventilation has been the most commonly employed treatment for combating Acute Respiratory Distress Syndrome (ARDS) [5]. Although Covid-19 is a novel disease, the virus has the ability to develop into a less familiar variant of the ARDS disease [2]. Thus, often when treating a patient with a severe case of Covid-19, the same treatments are employed as would be employed for an ARDS patient, i.e. mechanical ventilation. This is an invasive procedure that requires connecting the patient to the ventilator via a hollow tube, inserted into their mouth towards the primary airway or trachea. This procedure, due to its invasive nature, carries the risk of further infection or lung damage. The invasive nature of mechanical ventilation also makes it challenging to reduce the emission of aerosols from the patient, putting medical staff at risk. The World Health Organisation (WHO) guidelines on Covid-19 recommend High-Flow Nasal Oxygen (HFNO) therapy as a potentially effective and non-invasive alternative to mechanical ventilation [6]. Using the non-invasive method of supplying oxygen that is HFNO, rather
The use of a particle extraction vacuum visor for Covid-19 patients is proposed by this research and the research that has been undertaken to date in the University Hospital Galway and the National University of Ireland, Galway. The visor is to be secured to the head of the patient and can funnel exhaled particles into a vacuum tube, to be safely extracted from the environment in which the patient is being treated, as depicted in Fig. 1.

This study aims to design and optimise an existing visor design to better extract infectious particles from a patient undergoing HFNOT. The visor can be used without the need of a negative pressure isolation room because the visor itself is attached to a portable vacuum pressure suction tube to funnel infectious particles from the patient. Using high-fidelity approaches in computational modelling such as CFD simulation and advanced machine learning toolchains to facilitate the process of design and optimisation. Experimental testing was undertaken in order to validate both the CFD model and the abilities of the optimised visor design.

The methodology of the CFD model set-up and design optimisation process is detailed. The results are then analysed. A quantitative and qualitative comparison of CFD and experimental results is completed and conclusions drawn.

2. Description of methods

The aim of this research was to first assess the performance of an existing visor design (henceforth called the ‘original visor design’ in Fig. 1) using CFD and experimental methods. Then the visor design was optimised to maximise its infectious particles capturing efficiency. The design features to be optimised were the shape of the visor and a connection unit that attaches the tube to the visor. The visor performance efficiency was quantified by calculating the percentage of infectious droplets captured by the visor. This was validated experimentally using particle counting. The velocity profile of the droplets inside the visor and suction tube and any leakage of droplets from the gaps between the visor and the patient’s face was also analysed. A qualitative experimental validation of this analysis was carried through laser sheet visualisation and Schlieren imaging. The methods used in the CFD analysis, prototyping of the final designs and the experimental testing are documented in this section.

2.1. Computational domain, grid and other settings

The CAD model consisted of the patient’s face inside the visor and the visor attached to the suction tube (Fig. 2). First, the model was simulated with a suction tube connected to a visor without any connection unit. Then a design methodology was adapted to first optimise the shape of the visor and then optimise the connection unit, resulting in an optimised visor and tube assembly. For all simulations, the patient and visor model was enclosed inside a square domain of $0.8 \times 0.8 \times 0.8$ mm. This represented the surrounding environment outside the visor at atmospheric pressure – Fig. 2.

CFD can be used to model two scenarios, namely, breathing or coughing by the patient. Breathing will have a lesser mass flow rate and a lower velocity than coughing. The performance of the visor is therefore expected to reduce in a high droplet velocity coughing event, as the visor may be able to better capture low velocity droplets generated while breathing. The coughing event can be defined as a possible ‘failure’ event of the visor – as the visor is expected to capture low velocity droplets (breathing) better than high velocity droplets (coughing). Simulation modelling a patient’s cough was analysed in depth. Simulation modelling breathing was also carried out. However, an in-depth analysis of breathing simulation results will not be studied in this paper, as the coughing simulation was expected to simulate the worst-case ‘failure’ scenario.

This simulation uses two solvers simultaneously, namely, pressure based transient solver for continuous flow (air i.e., exhale) and transient Discrete Phase Model (DPM) solver for the discrete phase (water droplets i.e., saliva containing virions). Here, the discrete phase interacts with the continuous phase and the droplet trajectory is updated after each solver iteration. The mass flow rate of the continuous phase (cough and breath) was taken as 360 L/min and 6 L/min [13].

### Nomenclature

**Abbreviations**

| Abbreviation | Description |
|--------------|-------------|
| AMBU | Artificial Manual Breathing Unit |
| CAD | Computer Aided Design |
| CFD | Computational Fluid Dynamics |
| DPM | Discrete Phase Modelling |
| HFNOT | High Flow Nasal Oxygen Therapy |
| PPE | Personal Protective Equipment |
| PIV | Particle Image Velocimetry |
| WHO | World Health Organisation |

### Fig. 1. Mannequin receiving HFNO therapy with original visor design set-up.
In a cough the infectious droplets diameter vary in size. Thus, Rosin-Rammler droplet distribution was used with droplet diameter ranging from 1 µm to 100 µm.

The droplet range can be graphically fitted using Rosin-Rammler distribution law. For liquid sprays, Rosin-Rammler expression is a widely used and convenient representation of droplet size distribution [14]. Droplet diameter range of 1 – 100 µm was used [15] with a mean value of 80 µm [16]. Hence, the droplet diameter is not a singular value and is more accurately represented by a range. Droplets were generated at the mouth inlet. Droplets were modelled to behave accordingly when they collide with the model walls; (a) tube outlet captured the droplets, (b) square domain walls removed the droplets and (c) visor, tube inner walls and patient’s face reflected the droplets. Modelling surfaces as reflective is predicted to be an accurate assumption as the visor is not a porous material. Peng et al. suggests that when modelling droplets generated during coughing or sneezing, it is vital to incorporate friction coefficients of only porous media [17]. The droplets velocity was assumed to be same as that of the continuous phase (air) - 360 L/min. Air and water droplet evaporation was not considered in this simulation.

Air and water droplets were exhaled from the mouth, with suction pressure applied at the filter. Ambient atmospheric conditions were applied to the space between the visor and the face. An unstructured mesh was used and refined in regions of significant interest, i.e., around the mouth, face, visor walls, along the entrance of the inlet tube and at the inlet tube bend. Mesh convergence study was carried out and mass flow rate at the tube outlet was recorded. It was concluded that at least 460 K cells are needed for results to be independent of the mesh resolution – Fig. 3(c). Ansys Mesh was used to generate the mesh. The square domain has the dimensions 0.7 × 0.7 × 0.7 m. The average skewness of the entire model mesh is 0.223 and orthogonality is 0.771.

Detailed simulation settings are listed in Table 1.

### 2.2. Design optimisation approach

The overall goal in developing a structured approach to design is to be able to parameterise key dimensions or otherwise isolate discrete design inputs which can be iteratively optimised using CFD analysis. The key dimensions here is the shape of the visor and the design of the tube connection unit. First the original visor design is simulated and its performance evaluated by studying the air and water droplets velocities and trajectories. It is imperative to quantify the efficiency of different iterations of visors from the design optimisation study. A suitable metric that can be employed is the visor droplet capturing efficiency \( n \). This is the number of droplets extracted by the suction tube as a fraction of the droplets exhaled. The efficiency of the visor can be expressed as a percentage, where 100% is the most optimum visor design that captures all particles exhaled from the mouth.

Then adjoint solver is used to optimise the original visor shape. Parametric design method is employed to find the best-performing concepts and estimate their ideal dimensions, with the adjoint tool used to investigate whether further improvement may be possible. Geometrical changes in the visor suggested by the adjoint solver were used to reconstruct the CAD model of the visor. These geometries were then used in CFD simulations to calculate \( n \) and analyse the velocity and spread distribution of the air and droplets. Finally, the design of the tube connection unit was optimised following similar methodology. The optimised unit assembly containing the optimised visor and optimised connection unit was simulated in CFD.

Based on this design optimisation study, prototypes of the original and optimised visor were created to conduct CFD validation and experimental analysis — Schlieren imaging, laser
sheet visualisation and particle count tests. Fig. 4 summarises the optimisation process.

2.3. Prototyping

Rapid prototyping techniques, including 3D printing and thermoforming, were used to generate physical models of the CAD models, previously used in computational analysis. The original visor design and the optimised visor design were prototyped by thermoforming onto a 3D printed mould.

The mould was formed using VisiJet MR2-CL resin on a ProJet MJP 2500 printer. The printing material used was the VisiJet MR2-Cl resin. A secondary 3D printed part was generated that sits on top of the original inlet section of the original 3D printed mould to alter the base mould geometry in order to create the shape of the optimised visor design. This allowed for the optimised visor geometry to be thermoformed on top.

The original visor and the optimised visor design were created using a Formech 450DT thermoforming machine and hand-cut 350 × 550 mm, 1 mm thick Polyethylene Terephthalate Glycol (PETG) sheets. The Formech 450DT is a vacuum forming machine that allows for polymeric materials, such as PETG, to be heated and precisely moulded about a surface.

In both the attempts of prototyping the original and optimised visor, respectively, once the base mould or the underlaying two-part mould design were positioned, the PETG material was placed above the chamber, sealed and secured into position. This sealing is created by closing the frame about the chamber area and clamping about the chamber border. A heater is introduced to soften and heat the 1 mm thick PETG sheet. Through trial and error, it was determined that for the specific application required for this research, a heating time of 57 s was considered to be optimal. This optimal heating time is assuming that the machine has been preheated for approximately 15 min before use. Once the 57 s has elapsed, the heater is quickly retracted and removed from further interaction with the PETG. The base mould generated for the original design, or the two-part mould system (base mould and secondary conical part) for the optimised visor design, are raised into contact with the now ductile, workable plastic sheet. This causes the PETG to be reformed about the mould, taking the shape of the underlying visor design. Again, through trial and error, it was determined that the optimum time required for this process was 15 s. A vacuum is applied to the base of the bed for 5 s to allow for the PETG to be wrapped tightly about the mould, ensuring that all the required design features have been etched onto the PETG. These details of the thermoforming process are listed in Table 2. Finally, positive air pressure is applied to the bed once the PETG has cooled to allow for ease of separation of the base and the newly re-formed PETG sheet. The cooling period is essential to allow for the shaped PETG to establish geometric stability, ensuring that its removal does not cause unwanted features to be added.

2.4. Experimental setup and testing techniques

The experimental tests carried out are Schlieren imaging and laser sheet visualisation. The equipment setup for all three tests is depicted in Fig. 5.
A CPR manikin was used in the experimental setup, depicted in Fig. 5. The manikin’s nasal cavity allows for the humidified high flow vapour from the nasal passage to interact with the displaced airflow for the breathing or coughing experimentation. The CPR manikin was given a ‘trachea’ made of tubing which had Aerogen Solo nebulisers in line. The visor is secured using Velcro straps around the circumference of the forehead.

For breathing, an AMBU bag is used to displace air to simulate breathing at a rate of 12 breaths per minute. The AMBU bag delivers a max volume of 800 ml of displaced air to the system to simulate breathing when one hand stroke is used by the tester. This was considered acceptable for this simulation as the tidal volume of air inhaled and exhaled during normal breathing for an adult is 500 ml [18]. For the coughing simulations, the spray gun displayed in Fig. 4 was used to generate 12 coughs per minute, each at flowrates of 360 L/min in accordance with the CFD simulation completed. These experimental testing conditions are summarised in Table 3. A vacuum is attached to the outlet tube of the visor, which generated suction, to mimic the suction conditions generated by inbuilt hospital vacuums. Flowmeters were used to continuously check the flow rates being produced from the spray gun and from the vacuum tube.
Table 3

| Condition          | Value         |
|--------------------|---------------|
| Breaths per minute | 12            |
| Coughs per minute  | 12            |
| Vacuum flow rate   | 70 L/min      |
| HFNOT flow rate    | 60 L/min      |
| Coughing flow rate | 360 L/min     |

Fig. 5. Experimental test-rig setup for cough simulation.

Fig. 6. (a) Locations from which particle monitoring is conducted. (b) Gap created in surgical mask ‘seal’ due to HFNOT tubing.

2.4.1. Particle measurement

TSI SidePak AM510 personal particle monitor was used to quantify the particles escaping. Two methods were used in the experimental set-up to introduce particles and aerosols to the simulation during breathing simulations, namely, through HFNOT and from an Aerogen Solo nebulising device. The HFNO equipment provides humidified water vapour/aerosol to the manikin through nasal prongs, whereas the nebuliser generates an aerosol that is introduced to the system through the manikin’s trachea. This dual introduction of aerosol to the testing is conducted to simulate the actual volume of particles expelled during regular breathing – aerosol provided by the Aerogen Solo and to account for the increased volume of particles exhaled when HFNOT is used to treat patients – aerosol provided by an Armstrong Medical High Flow system. A third method of introducing particles to the setup is used in the cough simulations, with the spray gun vaporising a solution of distilled water and vegetable glycerin.

The locations from which particle monitoring was carried out are shown in Fig. 6, covering all sides of the face. Test data was gathered at the same 4 locations during the testing of the control (no visor), surgical mask and both original and optimised visor designs.

The particle concentrations were extracted at Location 2 as it has been found that this position is the main zone of aerosol leakage from a regular surgical mask due to the gap created by the HFNOT tubing. Location 4 is positioned at the chin of the manikin. It was predicted that this region will experience some aerosol leakage and therefore, must be monitored.

2.4.2. Schlieren imaging and laser sheet visualisation

Single mirror Schlieren techniques were employed to capture a visual depiction of the airflow from a manikin. A spherical, concave mirror of 300/1500 mm diameter was placed behind the manikin during the experiment. A LED light was used as a point source of light, shining in the direction of the mirror. A razor blade was used as the required sharp edge for this testing to block a percentage of the refracted light that is illuminated for the concave mirror. Both the sharp edge and LED source were positioned 2× the focal length from the mirror, in this case 2× 1500 mm from the concave mirror.

A Vision Research Phantom V310 camera with a Nikon Micro-Nikkor lens was used to record the results generated at high-speed. The components required for the Schlieren imaging and its layout is depicted in Fig. 7. Schlieren imaging was documented as the particle testing was being undertaken.

Laser sheet visualisation was conducted to have a better understanding of the airflow patterns that were being generated within the visor designs. The experimental test rig was setup as displayed in Fig. 5, with the addition of a DeWalt DCE089D1G-GB Green self-levelling multi-line laser level.

3. Results

The visor should be more efficient than existing devices at safely guiding droplets into the vacuum tube and reducing the number of droplets escaping into the atmosphere. If the escape of
a small number of droplets is inevitable, their velocity and spread must be reduced. The visor should be easily employed during HFNO treatment and comfortable to wear. The comfort wear of the visor was judged qualitatively, i.e. the visor should not be too large or protrude too far from the face. The visor should sit closely to the face.

The original visor geometry was simulated with a simple 90° elbow in the vacuum tube connected directly to the inlet. A velocity contour plot of the flow of exhaled air inside the original visor is presented in Fig. 8, showing the moment when exhaled air reaches the inlet of the vacuum tube.

The immediate concerns raised by Fig. 8 are that there is flow separation in the upper region inside the visor. This is caused by the high-velocity jet of exhaled air encountering the surface of the visor because of the inlet being positioned too low relative to the mouth. The two clearest areas for improvement in the design identified from this early stage were the geometry of the region of the visor closest to the inlet, which funnels exhaled air into the vacuum tube, and the position of the inlet itself relative to the mouth, which must be aligned sufficiently well to ensure that the visor does not obstruct the flow of droplets and aerosols into the tube.

3.1. Inlet region geometry - Concept analysis

When selecting the shape of the region of the visor geometry closest to the tube inlet (henceforth "the inlet region"), three general concepts were drawn, shown in Fig. 9.

The appeal of the conical design is that the distance between the inlet tube and the patient’s mouth can be easily extended, ensuring that the velocity at which the exhaled droplets encounter the visor is not unnecessarily high. The fact that the conical region slopes towards the tube inlet at all points is also beneficial as it means that it will tend to steer droplets directly towards the inlet. This is important also for meeting the aim of mitigating against droplets escaping at the rear of the visor.

The conical and hemispherical inlet concepts were compared using CFD analysis. For the conical design, the dimension varied is the height of the truncated cone (Fig. 10(a)), while for the hemispherical design it is the radius of curvature of the hemisphere that is varied (Fig. 10(b)).

Four different iterations of the cone height and the hemispheric radius were simulated in CFD and their particle capturing efficiency $n$ was calculated. This is graphed in Fig. 11.

The CFD results show that the conical designs tend to perform better than the hemispherical concept. Increasing the height of the cone tends to increase the efficiency of the design, though the marginal improvement drastically reduces once the height of the truncated cone exceeds a critical value of about 40 mm. This is a sensible result because it was predicted that increasing the distance between the mouth and the inlet would reduce the velocity of particles when the encounter the visor surface and thus reduce the rate at which they are deflected away from the inlet. It should be noted that the height of the cone is limited to about 40 mm. Further increase in cone height will protrude the cone further from the patient’s face. It is assumed that this will render the visor impractical and cumbersome to wear. For this reason, 40 mm height cone was the optimum design because the increase in efficiency achieved by a further 10 mm increase in height is less than 0.5% and does not justify the increase in length.

The performance of the hemispherical inlet concept is comparatively very poor, with only the 100 mm radius design outperforming the original visor geometry. This is likely because the distance between the tube inlet and mouth is not dramatically increased when the radius is altered. The key takeaway from these results is that the distance between the mouth and the inlet tube is the critical parameter in designing the inlet region of the visor.

3.2. Vacuum tube position analysis

The next stage of the design process was determining the ideal position of the vacuum tube inlet itself. A portion of the high-velocity jet emitted from the patient’s mouth impinges upon the interior surface of the visor just above the vacuum tube inlet. Elevating the position of the tube inlet relative to the mouth would greatly reduce the flow separation in the upper region of the visor and ensure that the exhaled jet with the greatest velocity will flow directly into the vacuum tube.

Analysis conducted using the adjoint solver concluded that raising the inlet of the vacuum tube would be the most effective
way to improve the efficiency of the design. The 40 mm conical design was analysed, with no changes other than the position of the inlet hole. The results are plotted in Fig. 12 where the visor efficiency is plotted against the perpendicular distance between the centrelines of the inlet hole and cone base.

A dramatic increase in efficiency is clearly achieved that is surprisingly close to approximating a linear relationship from 5 to 8 mm, before decreasing slightly when the inlet is raised further to 9 mm. The fact that the efficiency begins to decrease after an optimum value is unsurprising as the same flow separation due to excessive contact with the interior surface of the visor begins to occur below the tube opening if the hole is raised too much. This is further seen in Fig. 13(a) and (b).

3.3. Tube connection unit

The HFNOT vacuum tubes are built in or portable connection units in the hospital rooms, providing suction pressure from the hospital’s main supply. The vacuum tube should therefore, easily, and securely attach to the visor when in use and detached when HFNOT is completed. The design should ensure this flexibility. The tube insert has vanes, as a safety feature to avoid the potential for large objects to be sucked into the vacuum tubing.

3.3.1. Analysis of original tube connection unit design

The existing tube connection unit was first analysed. The front of the tube insert has six vanes of thickness 2 mm each, that attaches inside the visor outlet, shown by Fig. 14(b). The tube insert has six hooks around its circumference that latches on the ridges inside the tube, shown by Fig. 14(a).

It was expected that the vanes at the cross section of the tube insert would restrict the fluid flow in the tube. The velocity contour of the optimised visor geometry including the original tube insert is plotted in Fig. 15.

The optimised conical inlet shape of the optimised visor design allows the exhalation of fluid from the mouth to strike the centre of the tube insert. This is the desired direction of fluid flow; however, the design of the original tube insert does not allow for the entire plume to be captured in the vacuum tube. The excessive number of vanes and small diameter of the hole in the central region of the original tube insert causes obstructions and impedes the now streamlined cough plume. This causes the plume to be redirected back into the visor. The gap between the blunt edges of the insert and visor inner wall, highlighted in a red circle, further obstructs the flow path into the tube. Flow separation is observed, further explained by Fig. 16. The visor efficiency was calculated as 60%.

3.3.2. Geometry optimisation

Analyses of the CFD study completed with the original tube insert and optimised visor assembly, demonstrated a need to increase the central and outer diameter of the tube insert. The number of vanes needs to be reduced, to minimise obstructions in fluid flow path into the tube. The number of vanes were reduced from six to three of 1 mm thickness each. The central diameter and outer diameter were increased to 11 mm and 48 mm, respectively. CFD simulation of the modified insert was completed in Fig. 17 and compared with the initial design in Fig. 15.
Comparing Figs. 15 with 17(b), it can be shown that, greater amount of fluid flows through the tube insert that have three vanes as compared to six vanes. This result was expected as the three vanes provides a greater unobstructed cross-sectional area for the fluid flow, allowing more fluid to pass between the vanes. The modified insert in Fig. 17 optimises the blunt edges of the insert with more inclined edges (highlighted by a red circle), that creates a funnel opening of the insert, ensuring that the spread of the exhale leaving the mouth is contained and directed into the tube. This assembly was 66% efficient as compared to 60% efficiency of the original tube insert design.

3.3.3. Tube insert orientation analysis

The modified three vanes tube insert has rotational symmetry of order one. Hence, different rotational orientations of the tube insert can influence the droplet trajectories encountering the tube insert and the direction of droplet spread from the back side of the visor. This is predicted to influence the efficiency of the visor in mitigating the viral droplets spread. The four main different orientations of how the tube insert connects to the tube is shown in Fig. 18. It is assumed that the vertical vane in the first orientation is at 0°. This is rotated to 180°, 90° and 270° (or −90°), respectively.
Fig. 16. Front view of original tube insert.

Fig. 17. (a) Modified tube insert of 11 mm central diameter and 48 mm outer diameter. (b) Velocity contour plot of its respective airflow.

Fig. 18. Different orientations of the tube insert rotated at 0°, 180°, 90° and 270°.

As the vanes orientation can re-direct the fluid flow, critical regions for analysis are the tube inlet cross section and the rear end of the visor. Velocity contour was plotted to capture the plume escaping the visor, shown by Fig. 19. The white spheres denote droplets.

Velocity profiles are not symmetrical for 0° and 90° rotations. This is because the vertically unsymmetrical orientation of vanes inside the visor, redirects the path and direction of the fluid escaping from the rear end of the visor. Exhale escapes the visor from the entirety of the gap area between the face and the rear visor edges. This includes the chin region, head, and the sides.

When the tube insert with 180° rotation is used, exhale leaves the back side of the visor more uniformly. Fluid only escapes from the bottom of the visor near the chin area. Significantly less exhale escapes from the side of the visor. This is further explained by Fig. 20.

The droplet capturing efficiency of the visor with the different tube orientations were summarised in Table 4. The highest efficiency was observed when 180° rotation of the tube insert was used. An improvement of 3.73% is observed when compared to 90° and 270° rotations.

The central diameter was further increased to 52 mm from 48 mm. The outer diameter was increased to 18 mm from 11 mm. This allowed the insert to be form fitted in the visor, eliminating any gap between the tube insert outer inclined edges and visor inner walls. The tube insert was rotated to 180° based on the orientation optimisation study. The velocity contour plot is shown by Fig. 21.

From Fig. 21, it is evident that the majority of the exhaled plume is capable of flowing directly into the vacuum, extraction tube with less obstruction, in comparison to the flow separation displayed in the velocity contour plot of device with the initial (Fig. 15) and modified (Fig. 17) tube insert designs. The larger central diameter of the optimised tube insert allows for the unrestricted capture of most of the fluid flow into the tube, rather than deflecting it back into the visor. It allows a larger volume of fluid to travel down the tube, extracting particles, increasing the visor efficiency. The efficiency of the visor after implementing the optimisations analysed in this section, was increased to 73%.

3.4. Assembled visor

Fig. 22 draws a side-by-side comparison of the original and optimised assembly. Velocity contours were plotted at three-time history points.
Velocity contours taken at the tube cross section highlighted in Fig. 23(a), were plotted for both the original and optimised assemblies in Fig. 23(b) and (c) respectively.

A considerable improvement is observed in the optimised assembly's fluid flow profile when compared with that of the initial design. The velocity contour at tube cross section of the optimised design is more uniform and thus, indicates that the cough plumes being expelled is being almost directly captured by the vacuum tube, as the cough is originating directly in line with the tube's centreline (as shown in Fig. 22(f)).

A graphical depiction of the velocity profile of the tube is shown by Fig. 24. This data was extracted by drawing a line across tube cross section at Plane 1, location shown by Fig. 23(a).

It is evident that in the optimised design, the velocity profile spans the entirety of the tube cross section and the centre of the tube contains the highest velocities, meaning that almost the entire vacuum tube is being utilised to extract particles. It should be noted that the peak velocity of the exhale is observed at almost 50% distance of the tube, i.e., the tube mid-section, for the optimised visor. In comparison to this, for the original visor, peak velocity occurs at only 10% of the tube, rendering the remaining tube section as inefficient.

The next critical region for analysis is the rear end of the visor. This denotes the region where droplets and cough plume escape the visor, from the surrounding area between the face and the visor perimeter. This is presented in the velocity contour plots in Fig. 25, for both the original and optimised designs.

As indicated from Fig. 25(a), there is a much greater predicted spread of escaping airflow from the original design. Exhaled cough plumes escape from along the majority of the surrounding regions of the face, as indicated by the significant velocity that can be seen encompassing the face in Fig. 25(a). However, when the optimised configuration is used, the escaping droplet
Fig. 22. Velocity contour plots of (a) original visor and (b) optimised visor when fluid reaches tube insert. (c) original visor and (d) optimised visor when fluid reaches tube insert mid-section. (e) original visor and (f) optimised visor when fluid reaches suction tube outlet and fully develops. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Fig. 23. (a) Location of cross-sectional tube insert plane and subsequent velocity contours taken from (a) for the (b) original and (c) optimised visor designs.
Table 5
Particle measurement testing data from breathing simulation.
Breathing — Concentration (mg/m$^3$) over 2-min period

| Location 1 | Location 2 | Location 3 | Location 4 | Total   |
|------------|------------|------------|------------|---------|
| Surgical mask | 4.394      | 26.238     | 122.92     | 1.642   | 155.194 |
| Original visor design | 1.295      | 1.105      | 8.13       | 7.338   | 17.868  |
| Optimised visor design | 0.882      | 2.031      | 2.353      | 3.67    | 8.936   |

Table 6
Particle measurement testing data from coughing simulation.
Coughing — Concentration (mg/m$^3$) over 2-min period

| Location 1 | Location 2 | Location 3 | Location 4 | Total   |
|------------|------------|------------|------------|---------|
| Surgical mask | 1.323      | 5.625      | 3.131      | 1.898   | 11.977  |
| Original visor design | 1.073      | 1.113      | 2.798      | 1.159   | 6.143   |
| Optimised visor design | 0.504      | 0.853      | 1.851      | 0.909   | 4.117   |

Fig. 24. Velocity profile of the tube section for the original and optimised visor designs.

spread is reduced, shown in Fig. 25(b) only escaping from near the chin.

3.5. Experimental analysis

The results of the particle measurement testing are listed in Tables 5 and 6.

The total sum of the concentration of aerosol measured over a 2 min period in each of the 4 testing locations, were plotted in Fig. 26(a) and (b) for breathing and coughing experimental analysis, respectively.

The total aerosol escape from the surgical masks was significantly higher than those from either visor design. In the breathing analysis, in comparison to the surgical mask, the original and optimised visor designs allowed for 88% and 94% less aerosol escape into the environment, respectively. Furthermore, the total percentage improvement in aerosol extraction ability of the optimised visor design in comparison to the original was 50%.

In the coughing simulation, the original and optimised visor designs allowed for 49% and 66% less aerosol escape than the surgical mask, respectively. Comparing the visor designs, indicate that the optimised visor allowed for 33% less particle escape into the environment than the original when coughs are simulated. Thus, both visor designs have been proven to function significantly better than the current standard of preventing infectious particle spread, and the optimised visor proves to be superior to the original. Schlieren imaging results of the control (without any visor) are shown in Fig. 27.

Surgical masks have been documented to create gaps with the potential for particle leakage, on each side of a patients nose in the absence of a stiff nose bridge (Location 2), along with the leakage of particles from the gap created by the HFNOT tubing (Location 3). According to the particle testing data, these areas of concern permitted a significant amount of aerosol leakage to occur from the manikin wearing the surgical mask. 78.66% more aerosol escapes out from the side of the surgical mask, along the HFNOT tubing, than from the bridge of the nose during breathing. However, during coughing, the results in Table 5, indicate that the high-speed expulsion of a coughing event, leads to 44% more leakage from the bridge of the nose of the surgical mask in comparison to its side. This leakage of cough plume from a surgical mask is circled in Fig. 28(a) and (b).

Optimised visor design allowed for 71% less leakage from location 3 and 49.97% less leakage from location 4 than the original visor. This trend in improvement is also notable in the particle testing data measured during coughing, where the optimised visor permitted 33.845% less leakage from location 3 and 21.57% less leakage from location 4 than the original visor design. Fig. 29 displays both the original and optimised visor designs at the very beginning of a coughing event. The optimised visor design better contains the aerosol about the chin — Location 4, than the original visor design, where the aerosol immediately escapes into the environment (red circles). Furthermore, as highlighted by the white dashed circles, disturbances in airflow are immediately evident in the surroundings of the original visor, caused by the airflow escaping from the sides of the visor design. In comparison, the lack of disturbances in the surroundings of the optimised visor, and disturbances seen exclusively in the chin area, indicate that the optimised visor design has managed to contain the area of aerosol escape to a much more manageable, singular location — Location 4.

The intense grey colour that has been circled in Fig. 30(a) indicates a high concentration of aerosol escaping and significant disturbances in airflow from this location in the original visor design. Based on the disturbances in airflow about the chin — location 4 in Fig. 30(b), it can be concluded that the concentration of aerosol escape has been significantly reduced.

Laser sheet imaging results are shown by Figs. 31 and 32. The blue arrows in the images of the initial expiration of cough from the manikin in both visor designs Figs. 31(a) and 32(a), highlight the direction of the air flow. As highlighted by the red arrow...
in Fig. 31(b), the original visor design causes significant flow separation, and the aerosol plume flows back into the visor, rather than heading directly into the vacuum inlet. In the final stages of the analysis of the original visor design, as indicated by the blue arrow in Fig. 31(c), a stream of aerosol can be seen escaping down towards the chin region/location 4.
In Fig. 32(b), the increased visor inlet height allowed for the expiration from the mouth to be more in line with the centreline of the conical shape, causing less flow separation. Thus, towards the final stages of the breathing event in the optimised visor design, Fig. 32(c), an escaping stream of aerosol towards the chin area cannot be seen.

SAOImageDS9 software was used to generate better, contrasting visual images from the laser sheet visualisation results. In these images, a snapshot was taken just after the cough plume had made initial contact with the visor designs.

In Fig. 33, the dashed red lines indicate how far the visible flow separation for the expelled plumes reach within each visor design, following initial plume contact with the visor. The considerable small area between the broken red line, representing aerosol spread in Fig. 33(b), proves that the plume is directed towards the tube.

3.6. CFD and experimental analysis comparison

A summary of the design specifications applied to the optimised design to generate such improvement, are listed in Table 7.
Fig. 32. Laser sheeting of breathing simulation using optimised visor design (a) very start as airflow initially is expelled from the mouth (b) upon plume contact with visor and (c) towards end of exhalation. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Fig. 33. Enhanced laser sheet visualisation images of the spread of a plume from (a) the original visor design and (b) from the optimised visor design. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

Table 7
Optimised design specifications summarised.

| Visor                                            | Visor shape of outlet connection to tube | Cone configuration | Cone outlet diameter | 40 mm | Cone outlet location | 7 mm raised from the initial design (in-line with the mouth inlet) |
|--------------------------------------------------|----------------------------------------|--------------------|----------------------|-------|---------------------|------------------------------------------------------------------|
| Tube insert                                      | Number of vanes | 3                  | Vane thickness       | 1 mm  | Central diameter    | 18 mm                                                            |
|                                                  | Outer diameter | 52 mm              | Orientation          | 180° rotation |
|                                                  | Inlet diameter | 40 mm              | Bend                 | Upwards bend with internal ridges to allow latching on the insert. |
|                                                  | Length        | Dependent on vacuum connection set-up |

Some of the similarities between CFD and experimental results are qualitatively highlighted in Fig. 34).

The visor efficiency results calculated from experimental testing are compared to the abilities of a surgical mask to prevent the escape of aerosol in Table 8.

These values demonstrate significant improvement in the particle capturing abilities of both visor designs in comparison to the surgical mask, in both breathing and coughing simulations. This demonstrates the immediate benefit of replacing the current standard of protective equipment, the surgical mask, with this device, in the treatment of infectious patients. The final comparisons of the percentage efficiency improvement values made in this analysis are listed in Table 9.

A list of the overall improvements measured from CFD, and the main quantifiable enhancements measured through experimental methods, between the original and optimised visors, are compiled in Tables 10 and 11.

4. Discussion

From the design analysis completed it can be concluded that the visor with a conical shaped outlet region, is superior in mitigating viral droplets. The tube insert at 180° rotation with three vanes and central and outer diameters of 52 mm and 18 mm, respectively, is optimum. More flow separation is observed in Fig. 22(a) when the original visor is used, as compared to the
Table 10
Improvements made between original and optimised visor designs, determined by CFD.

| CFD optimisations made between original and optimum visor designs. |  |
|---|---|
| Overall efficiency — Breathing |  |
| Original visor | 46% |
| Optimised visor | 93% |
| Improvement | 47% |
| Overall efficiency — Coughing ‘failure event’ |  |
| Original visor | 35% |
| Optimised visor | 73% |
| Improvement | 38% |
| % Reduction in average velocity of any potential escaping droplets |  |
| Along sides of visor | 91% |

Table 11
Main quantifiable enhancements made between optimised and original visor design, found by experimental methods.

| Experimentally determined improvements made between the original visor and the optimised visor designs. |  |
|---|---|
| Particle testing data |  |
| Overall percentage improvement in particle extraction ability following a coughing event. | 33% |
| Percentage improvement in time required to manage aerosol post cough — how much faster the original visor is in ceasing aerosol escape into the environment, following a coughing event. | 45% |
| Schlieren imagery |  |
| Percentage reduction in distance of aerosol travel within the visor, caused by flow separation of aerosol plume, when plume initially contacts the visor. | 24.74% |

A main cause of flow separation by the original tube insert was its inclined edges creating a gap, highlighted by the black circle. This gap was eliminated in the optimised design by increasing the tube insert’s external diameter. The red circle highlight in Fig. 22(e) and (f), is a small gap where the tube latches on the insert’s hooks. This gap is dependent on the tolerances of individual components in the visor assembly. From Fig. 22(f), it is seen that the optimised visor ensures that the fluid flows between any gaps in the full visor assembly. The optimised visor design is therefore more efficient.

Fig. 22(c) illustrates that more of the exhale reaches the tube section when optimised visor is used. Since both visors velocity contours were plotted at the same fluid flow time in Fig. 22(c) and (d), it can be observed that more exhale enters the tube cross section in less time for the optimised visor, as compared to the original visor. Fig. 22(e) and (f) shows the fully developed cough flow. It is evident that the optimised visor captures majority of the exhale leaving the mouth, due to its larger central diameter and reduced number of vanes. Whereas, in the original visor, the exhale from the mouth is impeded by the tube insert reduced central diameter and six vanes. Hence, fluid fills the entire tube section in the optimised visor.

Significantly greater flow separation occurs in the initial design, causing large amount of exhaled plume to travel upwards inside the visor and leak from the back and sides. The conical geometry of the visor inlet region and the inclined edges of the tube insert funnels the flow into the tube, mitigating infectious spread. The widened central diameter of the insert captures the entire cough plume exhaled from the mouth. Whereas the original design proves to be inefficient in comparison due to the hemispherical visor region causing increased flow separation and impeding exhale flow into the tube due to the restricted opening of the original insert.

The particle testing and Schlieren imaging results discussed and presented, reflect the results generated by the CFD analysis. In both the CFD simulations and experimental tests, the same
mass flow rate of exhale and vacuum suction pressure were used, making the results comparable. The CFD results from the optimised visor design predicts a reduction in both overall number, and velocity of particles escaping from the back of the visor, with the highest speed particles being focused to the chin area — Location 4. This CFD evaluation is supported by the experimental analysis. CFD breathing simulations predict a 47% increase in efficiency and the experimental simulations measuring at a 50% improvement in visor efficiency. These compatible results, with merely a 3% deviation, are extremely satisfactory as this suggests that the CFD simulation can accurately replicate the experimental testing results. Furthermore, both the CFD and experimental simulations indicate that the optimised visor has increased in efficiency by 38% and 33%, respectively, in comparison to the original during coughing simulations. Again, a 5% deviation between the computational and experimental results is considered the tolerable, standard, acceptable variance. Some of the discussed similarities are highlighted in Fig. 32.

Through computational and experimental analysis, the authors propose an optimised shape of the visor to efficiently capture exhaled droplets. Further analysis can include evaluating the performance of the proposed optimised visor shape by conducting CFD and experimental analysis for a range of breathing and coughing droplet velocities. Droplet evaporation in the patient’s exhale can be incorporated in the CFD model.

5. Conclusions

A medical device to reduce the risk of transmission of contagious droplets from infected COVID-19 patients was introduced in this study. According to the CFD simulation carried out, the optimised visor design is 94% and 66% more efficient than a surgical mask during breathing and coughing events, respectively. It is 73% efficient in comparison to the 35% efficiency of the original visor design for a coughing event. Whilst breathing, the efficiency values are 46% and 93% for the original and optimised visor designs, respectively. Hence, the optimised visor improves 47% (breathing event) and 38% (coughing event) in preventing the escape of infectious particles. These are values in 5% and 3% deviation from experimental tests results. The main objective of this study, to generate a device capable of capturing a significant number of expelled particles from a patient, has been achieved. The optimised visor presented a better performance to extract particles from an infectious patient in a hospital setting, in comparison to both the original visor design and the current barrier method being used, such as surgical masks.

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.

Acknowledgements

This project is supported by the Discipline of Mechanical Engineering, College of Engineering and Informatics at National University of Ireland, Galway. This publication has emanated from research supported in part from the Translational Medical Device Lab. The authors gratefully acknowledge the continuous support, time and efforts of the Translational Medical Device Lab and the National Centre for Laser Applications.

References

[1] G. Spiteri, et al., First cases of coronavirus disease 2019 (COVID-19) in the WHO European region, 24 January to 21 2020, Eurosurveillance 25 (9) (2020) 2000178, http://dx.doi.org/10.2807/1560-7917.ES.2020.25.9.2000178.
[2] P.G. Gibson, L. Qin, S.H. Puah, COVID-19 acute respiratory distress syndrome (ARDS): clinical features and differences from typical pre-COVID-19 ARDS, Med. J. Aust. 213 (2) (2020) 54–56.e1, http://dx.doi.org/10.5694/mja2.506074, John Wiley and Sons Inc.
[3] W.H. Gan, J.W. Lim, D. Koh, Preventing intra-hospital infection and transmission of coronavirus disease in health-care workers, Saf. Health Work 11 (2) (2020) 241–243, http://dx.doi.org/10.1016/j.shaw.2020.03.001.
[4] B. Kennelly, et al., The COVID-19 pandemic in Ireland: An overview of the health service and economic policy response, Heal. Policy Technol. 9 (4) (2020) 419–429, http://dx.doi.org/10.1016/j.hlpt.2020.08.021.
[5] L. Gattinoni, M. Quintel, How ARDS should be treated, Crit. Care 20 (85) (2016) Accessed: Apr. 14, 2021, [Online]. Available: https://go.gale.com/ps/i.do?id=qHRCA&sid=googleScholar&linkaccess3fulltext.
[6] S. Leonard, L.I. Volakis, R. DeBells, A. Kahlon, S. Mayor, L. G. C. Dungan, Transmission assessment report: High velocity nasal insufflation (HVNI) therapy application in management of COVID-19, Sci. Innov. (2020) 1–25, VAPOTHERM, INC.,
[7] E.H. Sullivan, L.E. Gibson, L. Berra, M.G. Chang, E.A. Bittner, In-hospital airway management of COVID-19 patients, Crit. Care 24 (1) (2020) 292, http://dx.doi.org/10.1186/s13054-020-03018-x, BioMed Central.
[8] J.P. Frat, R. Coudray, N. Marjanovic, A.W. Thille, High-flow nasal oxygen therapy and noninvasive ventilation in the management of acute hypoxemic respiratory failure, Ann. Transl. Med. 5 (14) (2017) http://dx.doi.org/10.21037/atm.2017.06.52, AME Publishing Company.
[9] H. Library, HSE library guides: Covid-19 HSE clinical guidance and evidence: HSE operational pathway of care 006, 2021, Accessed: Apr. 14, 2021. [Online]. Available: https://hse.drstevenslibrary.ie/c.php?g=679077&p=486120.
[10] T.D. Girard, G.R. Bernard, Mechanical ventilation in ARDS; A state-of-the-art review, Chest 131 (3) (2007) 921–929, http://dx.doi.org/10.1378/chest.06-1515.
[11] A. Afshari, G. Hultmark, P.V. Nielsen, A. Maccarini, Ventilation system design and the coronavirus (COVID-19), Front. Built Environ. 7 (2021) 54, http://dx.doi.org/10.3389/FBUIL.2021.662489/BIBTEX.
[12] L. Moravskas, et al., How can airborne transmission of COVID-19 indoors be minimised? Environ. Int. 142 (2020) http://dx.doi.org/10.1016/j.envint.2020.105832.
[13] S. Hallert, J.V. Ashurst, Physiology, Tidal Volume, StatPearls Publishing, 2018, Accessed: Apr. 21, 2021. [Online]. Available: http://www.ncbi.nlm.nih.gov/pubmed/29494108.
[14] M. Alderliesten, Mean particle diameters. Part. VII. the Rosin-Rammler moment-ratio defined mean particle diameters, Part. Part. Syst. Charact. 30 (3) (2013) 244–257, http://dx.doi.org/10.1002/ppsc.201200021.
[15] X. Xie, Y. Li, H. Sun, L. Liu, Exhaled droplets due to talking and coughing, J. R. Soc. Interface 6 (SUPPL. 6) (2009) http://dx.doi.org/10.1098/rsif.2009.0388.focus.
[16] K.P. Fennelly, Particle sizes of infectious aerosols: implications for infection control, Lancet Respir. Med. 8 (9) (2020) 914–924, http://dx.doi.org/10.1016/S2213-2600(20)30323-4, Lancet Publishing Group.
[17] S. Peng, Q. Chen, E. Liu, The role of computational fluid dynamics tools on investigation of pathogen transmission: Prevention and control, Sci. Total Environ. 746 (2020) 142090, http://dx.doi.org/10.1016/j.scitotenv.2020.142090.
[18] V.C. Moore, Spirometry: Step by step, Breathe 8 (3) (2012) 233–240, http://dx.doi.org/10.1183/20734735.00217111.