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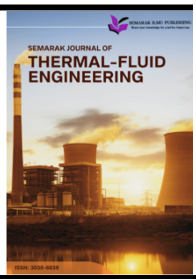

Modelling Analysis of Face Shield Effectiveness Against COVID-19 Transmission

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ABSTRACT

The ongoing COVID-19 pandemic necessitates effective protective measures to mitigate virus transmission. This study employed Computational Fluid Dynamics (CFD) to evaluate the efficacy of two face shield models in blocking COVID-19 transmission under three conditions: normal speech, coughing, and sneezing. One model replicates a common commercial product, and the other introduces an innovative design. A simplified human model with dimensions of 760 × 300 mm and a mouth air inlet area of 360 mm² was used for the simulation. Two types of face shields were modelled: one with a simple curved structure (Model 1) and another with a rectangular structure providing a side cover (Model 2). The computational domain was defined with dimensions of 3.5 m x 2.8 m x 2.3 m, and simulations were conducted using the finite volume method with ANSYS Meshing and Fluent for solver preference. The governing equations for the incompressible flow were applied. The simulations revealed that both face shields effectively blocked the direct airflow to the face across all conditions (speech, coughing, and sneezing). However, the structure of the face shields significantly alters the airflow patterns. Model 2, with its rectangular structure, provided better coverage and directed the airflow away from critical areas. Despite their effectiveness in blocking direct contact with airborne particles, face shields alone do not provide sufficient protection against virus transmission, especially for finer aerosol particles. Face shields can obstruct direct airflow but are inadequate as standalone protective measures against COVID-19 transmission. Therefore, the combined use of face shields and face masks is recommended for enhanced safety.

1. Introduction

The COVID-19 pandemic has profoundly impacted global populations, with infection rates surpassing the capacity of the healthcare systems in several countries [1-2]. Consequently, many infected individuals and patients with unrelated medical conditions are unable to receive adequate

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medical treatment [3]. To mitigate the spread of infection and maintain healthcare quality, it is crucial to implement effective preventive measures for both the medical staff and the general public [4]. Face shields, which cover the face with a clear plastic screen, along with medical masks, are widely used by healthcare workers to prevent inhalation of virus-laden droplets expelled through breathing, coughing, or sneezing by infected individuals [5-6].

Face shields have gained popularity as substitutes for masks in various settings, such as schools, universities, restaurants, and service industries due to their benefits. These include the visibility of facial expressions, ease of communication, reusability when properly sanitised, and increased comfort compared to traditional masks [7-9]. Studies using cough simulators have shown significant reduction in the risk of inhalational exposure when face shields are used [10, 11]. However, the effectiveness of face shields varies with the particle size and distance, offering less protection against smaller aerosols over extended periods [12, 13].

The use of face shields [14, 15] is not limited to healthcare settings; they are also used in various industries as part of personal protective equipment (PPE). Despite their widespread use, face shields lack standardised guidelines and their efficacy can vary based on design and usage [16, 17]. Previous studies have demonstrated that while face shields can block larger droplets [18, 19], they may not provide sufficient protection against finer aerosol particles, particularly when used alone without masks [20, 21].

Computational Fluid Dynamics (CFD) has been employed in previous studies to analyse the effectiveness of face shields. CFD simulations allow for a detailed examination of airflow patterns and droplet dispersion around face shields, providing insights into their protective capabilities under different conditions, such as normal speech, coughing, and sneezing. For example, face shields with various geometries have been modelled to assess how design modifications can influence their performance in directing airflow and preventing droplet ingress [22]. The airflow dynamics and particle dispersion were analysed by setting up boundary conditions that mimic human respiratory emissions. The mesh was carefully refined around critical areas, such as the mouth and edges of the face shields, to capture detailed flow features [23]. The simulations provided data on the airflow velocity, pressure distribution, and droplet trajectories, allowing for a comprehensive assessment of the effectiveness of the shields.

In this study, previous CFD methodologies were built to evaluate the efficacy of two different face shield models: one resembling a common commercial product, and the other featuring an innovative rectangular design with side coverage. Simulations were conducted using ANSYS Meshing and Fluent, applying the finite volume method to solve the governing equations for incompressible flow. This study aimed to determine the extent to which these face shields can block direct airflow and alter airflow patterns to enhance protection against COVID-19 transmission.

It was found that both face shields effectively blocked the direct airflow to the face across all tested conditions. However, the shield structure significantly affected the airflow patterns. The rectangular design (Model 2) demonstrated superior performance in redirecting the airflow away from critical areas, suggesting that design improvements can enhance the protective capabilities of face shields. Nonetheless, it has been reinforced that face shields alone are insufficient as standalone protective measures against COVID-19 transmission, particularly for finer aerosol particles. Therefore, combined use of face shields and masks is recommended for optimal protection.

2. Methodology

2.1 Geometry of Human Body and Face Shield

2.1.1 Human body

A simplified model of the human body was created for the simulation instead of using an actual human model. The dimensions of the human body were set to a height of 760 mm and width of 300 mm. The mouth of the human model, representing the air inlet, was modelled with an area of 360 mm², as shown in Figure 1.

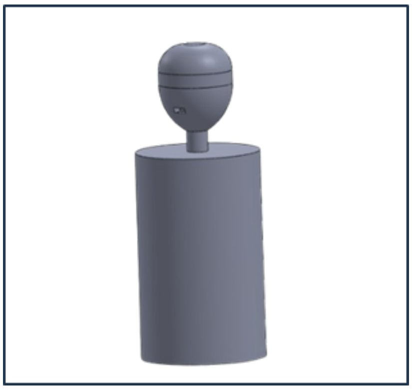
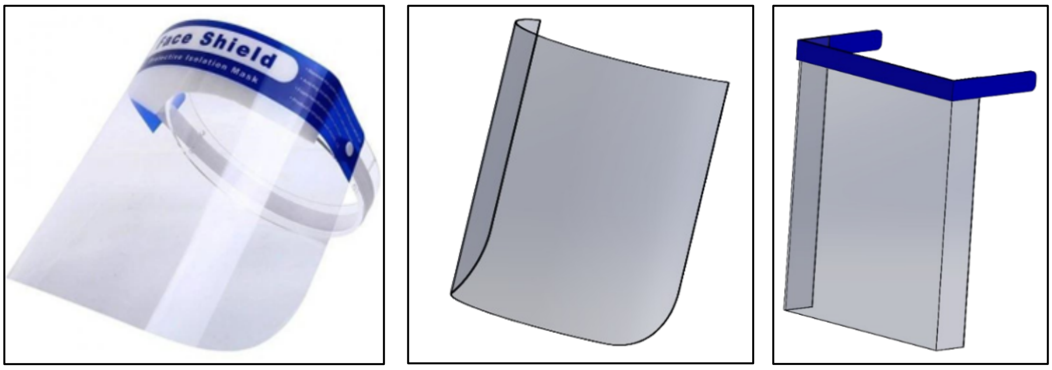


Fig. 1. Simplified human model

2.1.2 The geometry of the face shield

Two types of face shields were modelled for this simulation to test the effectiveness of different structures. The distance between the face shield and face was set to 25 mm. The commercial face shield shown in Figure 2(a) served as the baseline model. Face-shield computational fluid dynamics (CFD) models were created to simulate different structural designs. The first model (Model 1) is based on a commercially available face shield with a simple curved structure covering the face area. The headband was removed from the model to facilitate meshing. The second model (Model 2) had a rectangular structure designed to provide additional coverage to the sides of the face area. The detailed structures of these models are shown in Figure 2(b) and 2(c).



(c) (a) (b)

Fig. 2. Face shield CFD (a) Commercial face shield (b) Model 1 (c) Model 2

2.2 Computational Domain

The computational domain was meticulously defined to ensure accurate simulation of the effectiveness of face shields against COVID-19 transmission. The domain encompassed a space with dimensions of 3.5 m in length, 2.8 m in width, and 2.3 m in depth, providing a sufficient area to simulate human interactions while maintaining the recommended social distance of 1 meter between two human models, as shown in Figure 3.

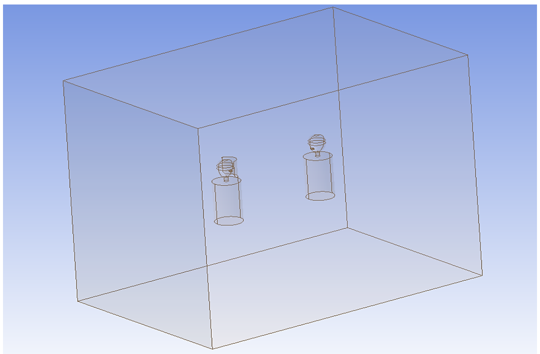


Fig. 3. Computational domain

2.3 Discretization Technique

The discretisation technique employed in this study involved the use of the finite-volume method. This approach facilitates the transformation of governing equations into a solvable algebraic form. ANSYS Meshing was utilised to create the computational grid, with the physics preference set for CFD and the solver preference set for Fluent. The initial mesh consisted of 68,000 nodes and 377,721 elements, predominantly tetrahedral in shape, as shown in Figure 4. To ensure the accuracy and stability of the simulations, grid independence tests were conducted, which involved adjusting the body-sizing parameters to achieve an optimal mesh density for the computational domain. By leveraging these advanced discretisation techniques, this study aimed to produce reliable and precise results regarding the efficacy of different face shield designs in mitigating the airborne transmission of COVID-19.

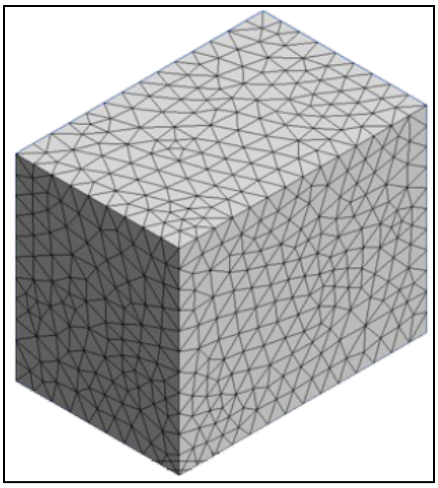


Fig. 4. Initial mesh for computational domain

2.4 Governing Equations

In this study, as referred to in the previous research equation [24], the fluid flow is considered incompressible, meaning that the density (ρ) remains constant. The governing equations used to

model the fluid dynamics are the incompressible continuity and the Navier-Stokes momentum equations. These equations describe the conservation of mass and momentum within the fluid domain, and are fundamental for understanding the flow characteristics around the face shields being evaluated.
Continuity equation:

$$\frac{\partial(u)}{\partial x}+\frac{\partial(v)}{\partial y}+\frac{\partial(w)}{\partial z}=0 \quad (1)$$

The continuity equation states that the sum of the rates of change of the velocity components (u, v, w) in the x, y, and z directions, respectively, must be zero, implying that the fluid is neither created nor destroyed within the flow field.

The momentum equations describe the conservation of momentum in the x, y, and z directions and account for the effects of pressure, viscous forces, and external forces, such as gravity. These are expressed as follows:

X direction Momentum:

$$u\frac{\partial u}{\partial x}+v\frac{\partial u}{\partial y}+w\frac{\partial v}{\partial z}=-\frac{1}{\rho}\frac{\partial p}{\partial x}+\gamma\left[\frac{\partial^2 u}{\partial x^2}+\frac{\partial^2 u}{\partial y^2}+\frac{\partial^2 u}{\partial z^2}\right] \quad (2)$$

Y direction Momentum:

$$u\frac{\partial v}{\partial x}+v\frac{\partial v}{\partial y}+w\frac{\partial v}{\partial z}=-\frac{1}{\rho}\frac{\partial p}{\partial y}+\gamma\left[\frac{\partial^2 v}{\partial x^2}+\frac{\partial^2 v}{\partial y^2}+\frac{\partial^2 v}{\partial z^2}\right]-\rho g \quad (3)$$

Z direction Momentum:

$$u\frac{\partial w}{\partial x}+v\frac{\partial w}{\partial y}+w\frac{\partial w}{\partial z}=-\frac{1}{\rho}\frac{\partial p}{\partial z}+\gamma\left[\frac{\partial^2 w}{\partial x^2}+\frac{\partial^2 w}{\partial y^2}+\frac{\partial^2 w}{\partial z^2}\right] \quad (4)$$

These governing equations were discretised and solved numerically using the finite volume method to simulate the airflow around the face shields under different conditions. This approach allows for a detailed analysis of the flow characteristics and effectiveness of face shield designs for mitigating the transmission of airborne particles.

2.5 Boundary Conditions and Parameter Assumptions

In this study, three conditions were simulated to evaluate the effectiveness of face shields in different scenarios. These conditions were applied equally to both the human models within the computational domain. Based on Figures 3 and 5, the individual wearing the face shield was positioned on the left side, whereas the individual not wearing the face shield was positioned on the right side of the computational domain.

The boundary conditions were set as the velocity inlet was established at the mouth surface for both individuals, simulating respiratory emissions. The pressure outlet was defined at the right wall of the computational domain, allowing airflow to exit the simulation space. The surfaces of both human models and the face shield were treated as non-slip walls to simulate the interaction between

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the airflow and physical barriers accurately. Although a transient study with a respiratory velocity profile would yield more accurate results, this study assumed a steady-state simulation. Thus, the maximum velocity from the respiratory velocity profile was used as an absolute value for the velocity inlet.

By incorporating these boundary conditions and assumptions, this study aimed to provide a realistic simulation of human respiration and its interaction with face shields, focusing on the effectiveness of these shields in different respiratory scenarios, such as normal talking, coughing, and sneezing.

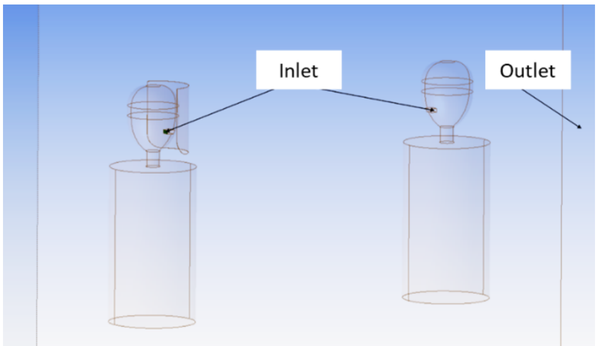


Fig. 5. Inlet and outlet

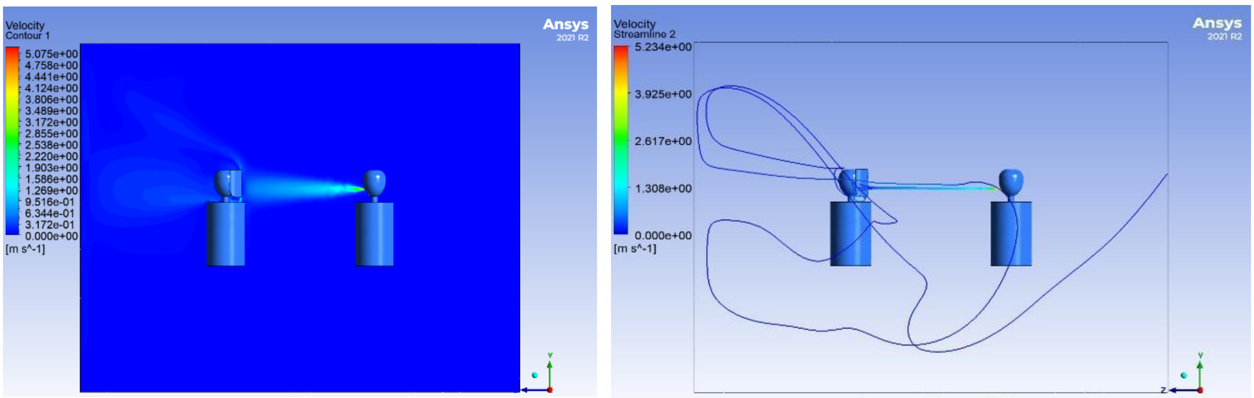
3. Results

3.1 Face Shield CFD Model 1

3.1.1 Normal talk condition

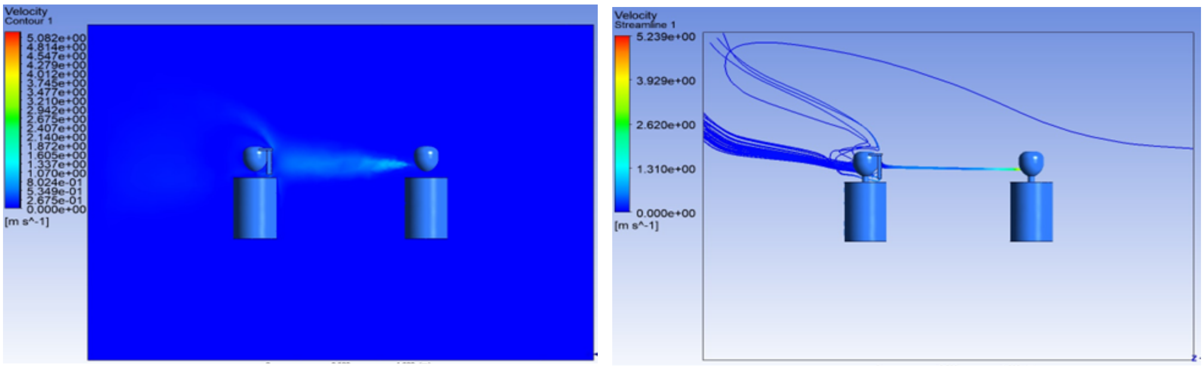
In Model 1, during normal talk, the highest velocity observed was 5.075 m/s at the mouth, as shown in Figure 6. The airflow from the person without the face shield was blocked by the face shield worn by the other person. The generated flow was deflected by the shield, primarily towards the back, top, or bottom sides of the shield. The velocity streamlines were minimal and non-fibrous owing to the low velocity.

In Model 2, as shown in Figure 7, the velocity during the normal talk was 5 m/s. Similar to Model 1, the face shield effectively protected both individuals from each other's mouth flow, with streamlines being deflected back towards the person wearing the shield. Comparing the two models, Model 2 showed a marginal improvement of approximately 1.5% in deflecting the airflow away from the critical face area.



(a) (b)

Fig. 6. Velocity (a) Contour (b) Streamline for normal condition of Model 1



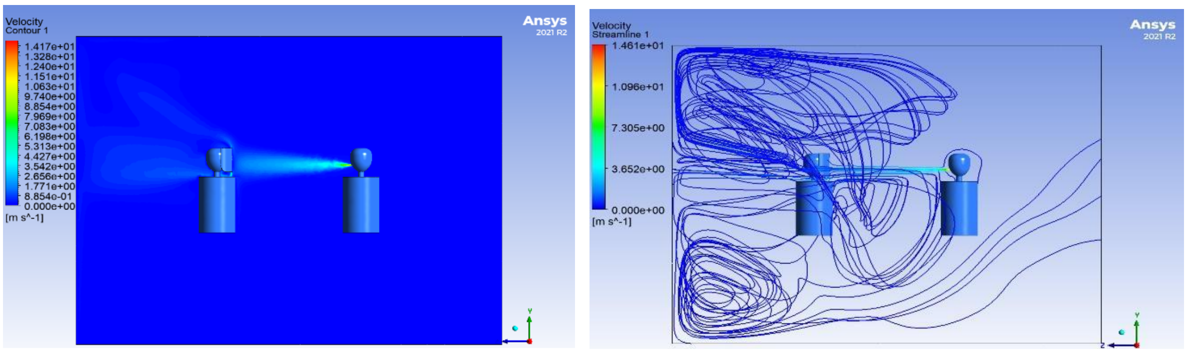
(a) (b)

Fig. 7. Velocity (a) Contour and (b) Streamline for normal condition of Model 2

3.1.2 Coughing

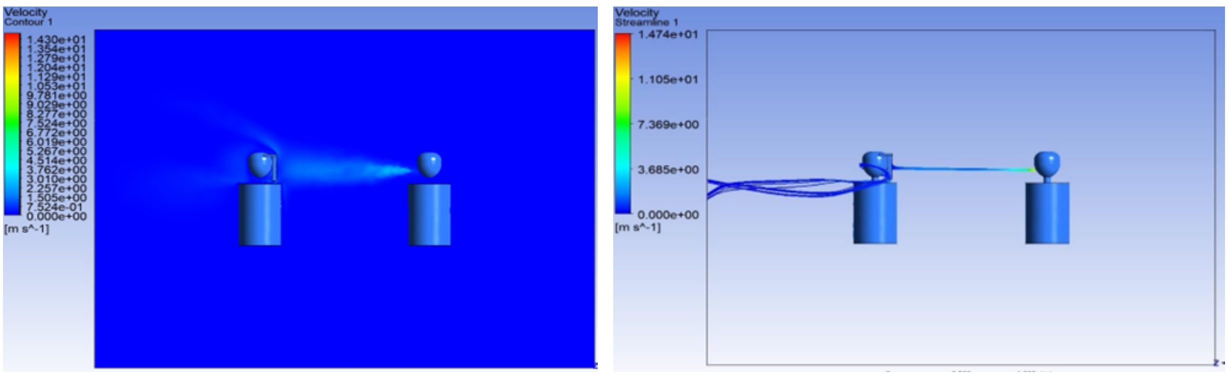
During coughing, Model 1 exhibited a peak velocity of 14.17 m/s at the mouth, as illustrated in Figure 8. The face shield effectively blocked airflow and prevented direct contact with the face area. The velocity streamlines were higher and more fibrous compared with normal talking, with a notable deflection on the left side of the domain.

In Model 2, shown in Figure 9, at a velocity of 14.2 m/s, the airflow from the second individual significantly impacted the face shield of the first individual, but was deflected around the face. The face shield effectively blocked the airflow, ensuring that the second individual was protected from direct exposure. Model 2 showed a slight improvement of 0.2% in reducing the direct airflow contact compared with Model 1. However, the overall effectiveness remained similar, with both models exhibiting significant airflow deflections.



(a) (b)

Fig. 8. Velocity (a) Contour (b) Streamline for coughing condition of Model 1



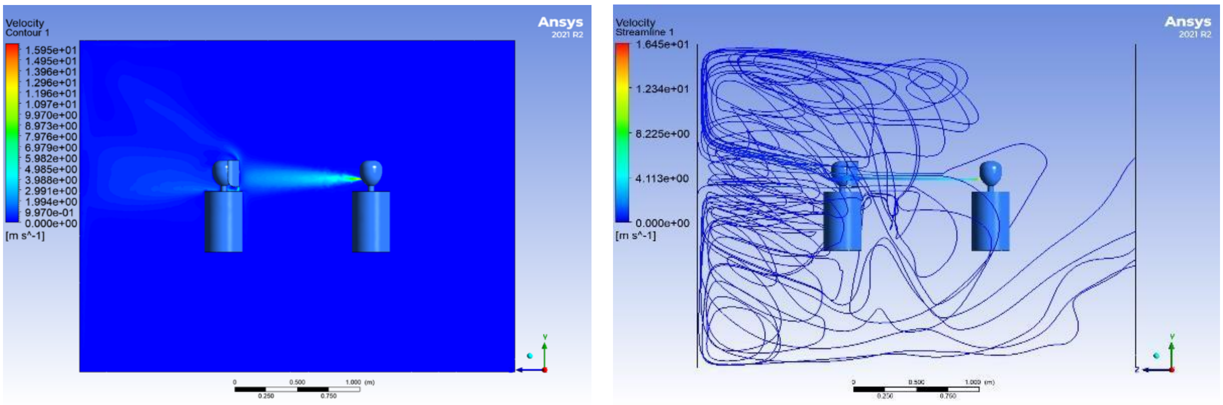
(a) (b)

Fig. 9. Velocity (a) Contour (b) Streamline for coughing condition of Model 2

3.1.3 Sneezing

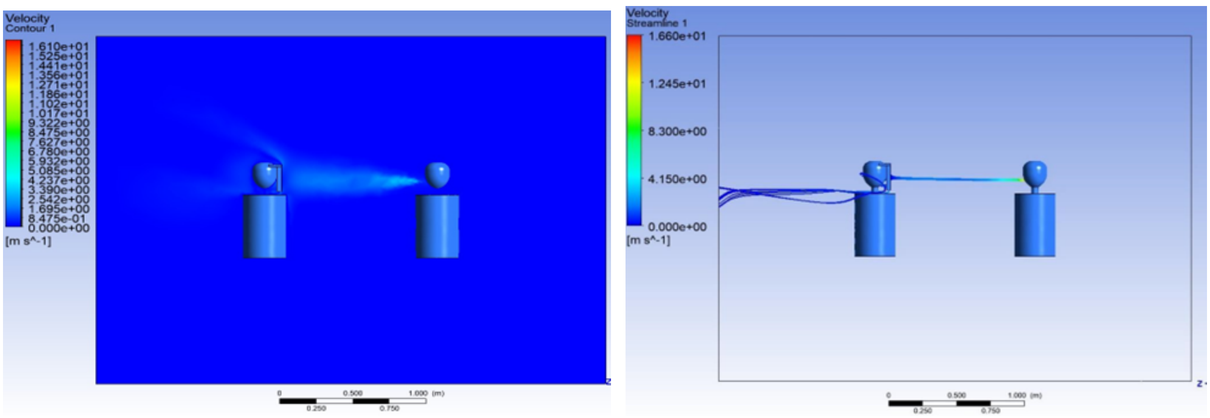
For sneezing in Model 1, the highest recorded velocity was 15.95 m/s at the mouth, as shown in Figure 10. The velocity streamlines were similar to those observed during coughing, but with higher intensity, concentrated towards the left side due to deflection by the face shield.

In Model 2, sneezing at 16 m/s showed the highest velocity recorded in this study (Figure 11). The face shield effectively blocked the direct flow from the second person, with streamlines deflected back towards the person wearing the shield, preventing the flow from reaching the second individual. Compared to Model 1, Model 2 exhibited a minor improvement of 0.3% in handling the high-velocity airflow generated during sneezing.



(a) (b)

Fig. 10. Velocity (a) Contour (b) Streamline for the sneezing condition of Model 1



(a) (b)

Fig. 11. Velocity (a) Contour **(**b) Streamline for the sneezing condition of Model 2

3.1.4 Discussion on comparative analysis between Model 1 and Model 2

Model 1, as shown in Figure 12, demonstrated that normal talking generated lower airflow velocities compared to coughing and sneezing, with the latter two showing similar velocity contours. The face shield was effective in blocking airflow within a safe distance under all conditions, although normal talking was deemed safer because of the lower velocities and fewer streamlines. However, for coughing and sneezing, the face shield alone was not sufficiently effective because of the higher velocities and fibrous streamlines concentrated around the person wearing the shield, as shown in Figure 13.

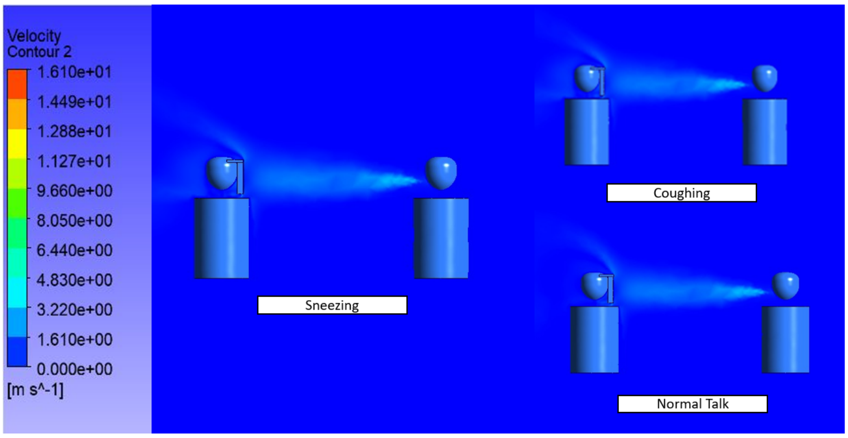


Fig. 12. Comparison of velocity contour for Model 1

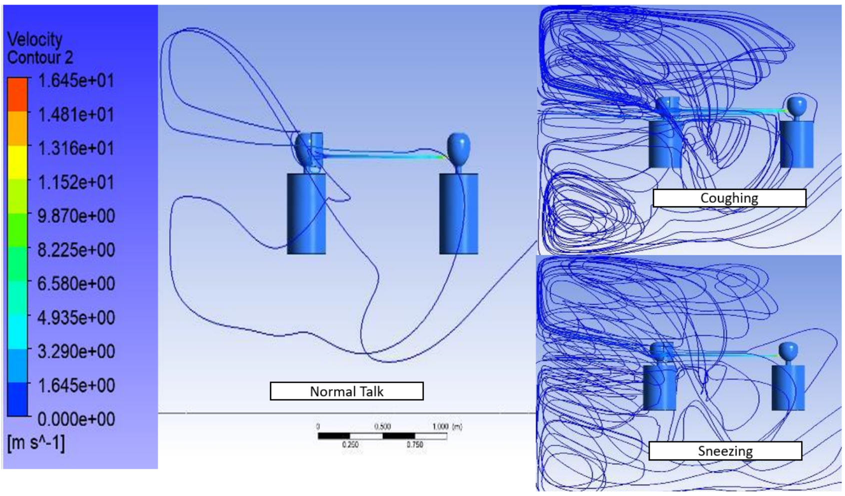


Fig. 13. Comparison of velocity streamline for Model 1

Model 2 results indicated that sneezing had the most significant impact on the face shield, with the highest velocity, as shown in Figure 14, and more pronounced effects than normal talking and coughing. The face shield successfully deflected the airflow, but in an enclosed space, reversed flows posed a risk. Overall, the face shield provided protection, but was more effective for normal talking than for coughing or sneezing because of the lower airflow velocities involved in normal talking. The velocity distributions and streamlines are shown in Figure 15.

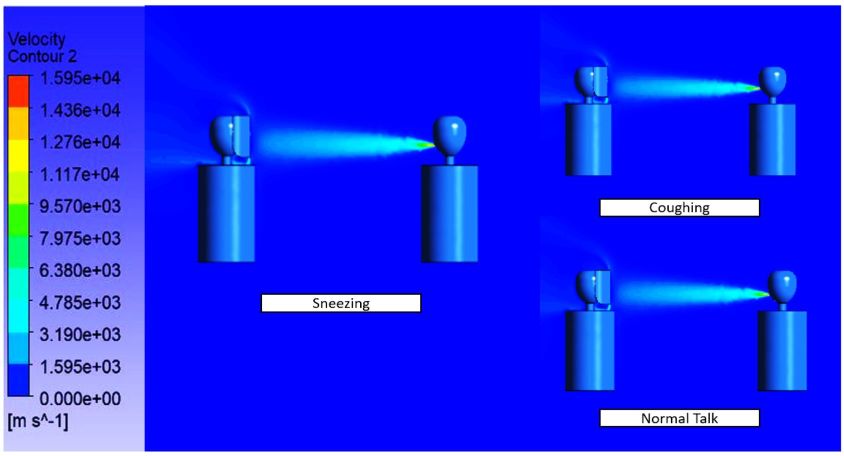


Fig. 14. Comparison of velocity contour for Model 2

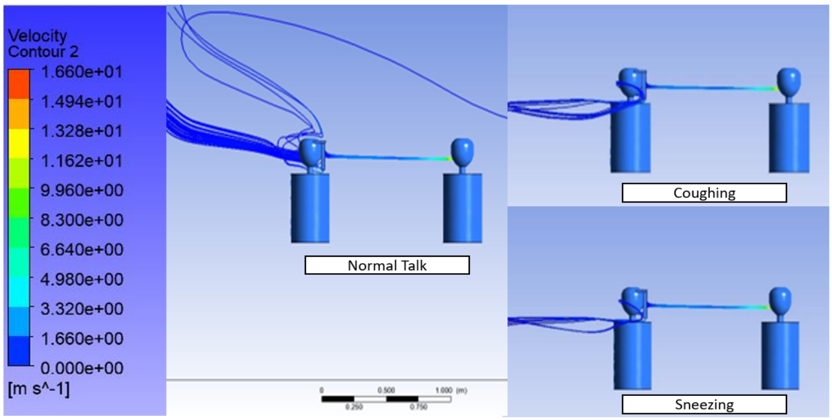


Fig. 15. Comparison of velocity streamlines for Model 2

Overall, Model 2 showed marginal improvements in protecting against airflow compared to Model 1 across all scenarios. Both face shields were effective in blocking direct airflow to the face, with Model 2 offering slight enhancements in deflection and protection. However, the differences between the two models were relatively small, indicating that while design improvements can enhance protection, both models still provided substantial defense against normal talking, coughing, and sneezing. The combined use of face shields and masks remains recommended for optimal protection, as shown in Figure 13.

As seen in Figure 16, the highest velocity magnitude occurs at the position where the airflow generated by both models comes into contact with the face shields. It can be concluded that face shields are considered more effective in normal talking conditions but not effective enough for coughing and sneezing.

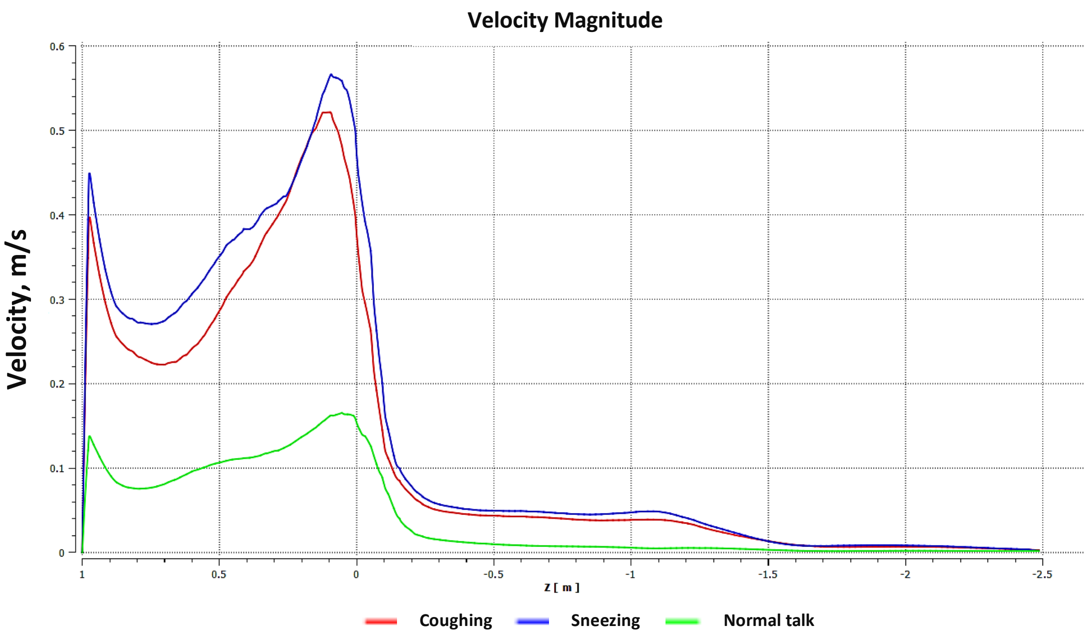


Fig. 16. The velocity magnitude

4. Conclusions

In conclusion, Model 2 demonstrated superior performance compared with Model 1 in several key areas. It showed a 1.5% improvement in deflecting airflow away from the critical face area, a 0.2% enhancement in reducing direct airflow contact, and a 0.3% increase in managing the high-velocity airflow generated during sneezing. These findings suggest that Model 2 may provide improved protection in environments in which airflow control is crucial.

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