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Flow Visualization and Aerosol Characterization of Respiratory Jets Exhaled from a Mannequin Simulator

by

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A thesis submitted in partial fulfillment of the requirements for the degree of Master of Science in Mechanical Engineering
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Abstract

It is important to understand the airborne transmission of infectious diseases due to the COVID-19 pandemic. Physical experiments were conducted to understand the dispersion of aerosols emitted from a mannequin simulator replicating human tidal breathing through the nose and mouth with and without a protective face mask in an enclosed space. Flow patterns are observed via high-speed visualization, and the concentration and size distribution of the particles is measured as a function of distance from the mannequin using an optical particle sizer. Masks were shown to effectively reduce the horizontal dispersion of aerosol for both mouth and nose breathing. Further, the masks were effective in decreasing the concentration of emitted aerosols and were particularly effective for the largest particles. The results provide insight into the dispersion of aerosol by breathing and into how masks may help contain that aerosol to prevent the spread of airborne diseases.
Chapter 1 Introduction and Literature Review

1.1 Background on Sars-Cov-2

Transmission of airborne diseases has been an interest of issue around the world in the past and more so in the present. Several cases of pneumonia and its unknown origin reported in Wuhan city of China in 2019 lead to a global pandemic. The cause of these cases was later identified as coronavirus, abbreviated as Sars-Cov-2 by the World Health Organization. This pathogen was initially identified by samples collected from throat and nose among the infected and is found to have a severe impact on the respiratory system. Due to the unknown mode of transmission and its high mortality rate, WHO declared Covid-19 as a public health emergency of international concern [26].

Many studies and experiments are being conducted by scientists, academicians, researchers, and medical professionals around the world to understand the origin and transmission of SARS-Cov-2. It is crucial to recognize and prevent the spread of the virus as it has affected more than 150 million people with a death toll over 3 million worldwide. A few studies reveal SARS-Cov-2 to be a novel human-infecting Betacoronavirus. There are some similarities to deadly pathogens like SARS associated coronavirus dated back to 2003 epidemic, which was first identified in Asia and spread to a dozen countries before it was contained. Genome analysis of SARs-Cov-2 indicates similarities to bat-derived coronavirus that was previously detected in Mojiang miners’ case in Yunnan province of China in 2012 due to horseshoe bats (Rhinolophus) [24].
Respiratory infectious diseases like the common cold, influenza, and tuberculosis are often transmitted through air. SARs-Cov-2 is also an airborne disease which can spread directly through aerosol particles and indirectly via surfaces [18]. It is believed through extensive research; airborne organisms transmit through air with the particles that create a feasible home for them and could exist in the surroundings long enough to infect people or animals. It is very important to sanitize and maintain health facilities as they are more prone to transmission of infectious diseases.

1.2  Aerosols and Its Behavior

Aerosol comprises solid or liquid particles suspended in any gaseous medium. They are present in several forms like bioaerosols, dust, cloud, fume, haze, mist, and smog. Bioaerosols are airborne particles that carry viruses, bacteria, fungi, pollen, etc. The micro-organisms can be the pathogens present in the dust particles that are infectious. Inhalation of these pathogens by human beings and/or animals can lead to severe respiratory complications and rapid health effects [25].

Particles are either generated naturally or are biogenic and emitted directly into the atmosphere (primary aerosols) or by chemical reactions (secondary aerosols) in the respective medium. It is important to understand the physical and chemical composition of aerosols and their behavior in the environment as these factors influence the properties and transmission of the particles. The size distribution, shape, density, mass concentration, and travel distance play important roles in transmission and will be briefly reviewed.

1.2.1  Particle Size

The size of a particle plays a key role in determining the behavior of an aerosol. Atmospheric aerosols that we breathe constitutes for polydisperse aerosols due to wide-ranging size and different mechanisms of generation. Whereas monodispersed particles are of equal size range and produced in laboratories as test aerosols. Particle size distribution of polydisperse
aerosols help in characterizing the nature of particles. These distributions can be represented graphically and mathematically based on probability functions and log functions.

Small particles can travel farther distances, stay on a surface for a longer period, and can transmit disease through airborne route. The smallest droplets often have the highest pathogen concentration because of evaporation. In contrast, large particles succumb to environmental changes such as gravitational force, humidity, evaporation, condensation, and diffusion transmitting via droplet or contact routes [13]. Large particles undergo ballistic trajectories due to gravitational and aerodynamic drag. Mechanism of aerosol generation, liquid content and viscosity of the aerosolized fluid influence the particle size.

1.2.2 Particle Shape

The aerodynamics and diffusivity of an aerosol are affected by the shape of a particle. The shape of a particle depends on the source of formation and the changes after forming due to crystallization, hydration, agglomeration, etc. Many particles are spherical in shape due to surface tension. In complex shaped particles, the equivalent volume is considered to obtain a representative droplet diameter. Since the transport behavior of irregularly shaped particles is difficult to express mathematically, in some cases a dynamic shape factor is considered, which is the ratio of resistance of a given particle to that of a spherical particle having the same volume. [25]

1.2.3 Particle Density and Concentration

The chemical composition of aerosols is extremely varied. The particle density remains constant throughout the transmission and changes only if it undergoes oxidation or hydration. Particle density is the mass per unit volume of a substance which is different from that of concentration. Density of the particle changes with change in the concentration of a solution. Solid and liquid particles have densities equal to parent material whereas smoke or fume particles have
densities less than the intended chemical composition. They are measured as kg/m³ or g/cm³. Concentration is the mass of solute per volume of solution (amount of substance in another substance). Concentration of an aerosol is the mass of particulate matter in unit volume of an aerosol. Number concentration is a common measurement known as number of particles per unit volume of aerosol. It is commonly expressed as number/cm³ or number/m³.

1.3 Important Factors of Aerosols and Airflow

Understanding the theory behind interactions between the particles and the suspended medium (gas) is important in transmission of the aerosols. The surrounding gas affects the particle a great deal when interacting with them. Temperature, pressure, mean free path and viscosity of a gaseous medium has an impact on the motion of suspended molecules and will be discussed in the following sections.

Brownian motion is the motion of a particle that takes place in still air. Diffusion of particles takes place from higher to lower concentration gradient. Particle motion is very simple for small aerosols under Brownian force. Due to their size, gravity and particle moment are negligible and impulsive motions due to collisions are damped due to drag force. Puffs and jets are driven by momentum while plumes and thermals are driven by buoyancy. The flow moves forward and away from the source with increasing distance.

Molecular velocity increases with an increase in temperature of the gas and can be calculated by a combination of ideal gas law and Boyle’s law known as rms (root mean square) velocity. When molecules are suspended in air, they tend to collide with each other or the walls of confinement and hence measuring the distance between each collision known as mean free path is essential. This distance is directly proportional to the temperature and inversely proportional to pressure.
Viscosity and diffusivity of gas are important factors to be considered for a particle motion, where viscosity increases with rise in temperature in gaseous medium but contradicts the theory in liquid mediums due to cohesive forces. These forces usually decrease with rise in temperature and are not significant as the gas molecules are spaced far from each other. Breathing, coughing, and sneezing in a human can cause tremendous initial particle velocities.

Flow dynamics might not affect large droplets as it does small droplets. For example, a turbulent cloud could have circulatory flow and move upwards due to buoyancy. The buoyancy is caused by density difference and temperature. Small droplets will be carried upwards with the flow whereas larger droplets will likely be left behind. The rate of evaporation of a droplet depends on the difference between the droplet surface saturation vapor pressure and the vapor pressure of the surrounding air depending on humidity. It also depends on mass diffusion coefficient and relative velocity making the Reynolds, Nusselt, and Sherwood numbers important nondimensional parameters. Reynolds Number \( R_e \) helps in determining the flow pattern of a fluid either to be laminar or turbulent. At low Reynolds number, the flow leans towards laminar flow dominating viscous forces whereas at high numbers, the flow tends to be turbulent dominating inertial forces. Turbulence occurs because of variation in fluid speed and direction (causing eddy currents and vortices). Fluid friction plays a role in developing turbulent flow. \( R_e \) determines the transition of flow from laminar to turbulent. The equation to calculate Reynolds Number is

\[
R_e = \frac{\rho u L}{\mu}
\]

where \( \rho \) is the density of the fluid (kg/m\(^3\)), \( u \) is the flow speed (m/s), \( L \) is linear dimension (m) and \( \mu \) is the viscosity (Ns/m\(^2\)). Defined as the ratio of inertial viscous forces.
Stokes law is the general differential equation over time to determine the behavior of fluid motion, derived from Newton's second law of motion. At low Reynolds number, stokes law can be used to calculate the viscosity of the fluid and fall velocity of the particle. The speed of the particle also decreases over time due to evaporation.

1.4 Environmental Effects on a Droplet

Molecular transfer from the particle to the surrounding gas results in nucleation, evaporation, and condensation. Vapor pressure, which is directly related to concentration of vapor in volume of gas and the temperature given at a point, play a role in the above-mentioned processes. The saturation ratio, which is defined as the ratio of partial vapor pressure of the system to that of saturation vapor pressure at a given temperature, must be calculated for further understanding. Vapor to particle formation by condensation is called nucleation. The size of a particle increases with condensation and decreases with evaporation of vapor. These effects can be estimated by gathering information about the surrounding air temperature and relative humidity.

The role of humidity plays an important role in survival of the pathogen lead particles in closed spaces. Relative humidity (RH) also known as saturation ratio, when expressed in percentage is 100 times that of saturation ratio. Aerosols either grow or evaporate when expelled into the surrounding depending on the temperature and relative humidity. When the evaporation of a particle takes place, due to water loss, the infectious concentration in these shrunken particles increases and remains in air for a longer period. This leads to increase in contamination levels. From one of the studies conducted in recent times, it was found that the virus viability is higher at RH <40 and >100. At a lower RH, rapid evaporation dried out the droplet and at higher RH, the droplet did not dry out completely and could still carry pathogens increasing solute concentration and viability of virus in the surroundings. [25&26]
1.5 Aerosol Generation and Measurements

Generation of aerosols may occur via biological processes such as coughing or breathing or through medical procedures. Both generation methods will be briefly reviewed. The aerosol generation and transmission process has several stages which constitute complex flow phenomena such as air–mucous interaction, liquid sheet fragmentation, turbulent jets, and droplet evaporation and deposition, to flow-induced particle dispersion and sedimentation, thus making flow physics the center of the process. The expulsion of droplets and the splatter pattern through nose and mouth when breathing, coughing, and sneezing play a key role in transmission of infectious diseases from one individual to another. In addition, micro-organisms present in the particles originate from animal or human feces and are present in ambient air. These pathogen-carrying particles can be suspended in air when disturbed by air flow and human activities.

It is believed that the higher viral load of SARS COVID virus is present in upper respiratory tract of asymptomatic hosts who produce virus laden droplets during normal activities such as coughing, breathing, and speaking. The droplet formation takes place due to two key mechanisms in a human. One mechanism is the instability and eventual fragmentation of the mucus lining due to the shear stress induced by the airflow. The other mechanism is due to rupture of fluid lining when the respiratory passage is opening [7]. Salivary droplets due to contact of tongue and lips is another mechanism to be noted. During a cough, mucus and other fluid secretions in the airways and the lungs get cleared. These droplets are expelled into the air at high velocities.

Based on a study conducted by Bourouiba et al [2], the transmission of the infected aerosols can be categorized into 3 modes: Direct contact with bodily fluids, large droplet transmission by spraying the infected particles in/onto conjunctiva or mucus lining, and airborne transmission via small droplets that remain in air longer and can travel farther. These small droplets usually
evaporate and form droplet nuclei which is the solid residue. An expulsion jet due to cough or sneeze can cause turbulent flow consisting of buoyant hot moist air and droplets ranging across various sizes. The evaporation and settling time of a droplet depends on its size. Droplets with diameter \( d > 100 \, \mu\text{m} \) have settling times less than 1 s whereas droplets with diameters of \( d < 5-10 \, \mu\text{m} \) have settling speeds of 3 mm s\(^{-1}\) and may take many minutes to settle to the ground. In addition, these small droplets can be resuspended due to flow movements or any external air source (e.g., air conditioning in enclosed spaces). In addition, the critical size that differentiates large to small droplets can be affected by temperature and humidity variations [2].

Morowska et al (2008) measured the expiration air jet velocity of human coughing and speaking and further characterized the size and concentration count of the expiratory droplets at the mouth using particle image velocimetry (PIV) and interferometric Mie imaging (IMI) [6]. These authors found that coughing had an average speed of 11.7 m/s while speaking generated an average speed of 3.9 m/s. These authors also found that the geometric mean diameters of droplets generated by coughing and speaking were 13.5 \( \mu\text{m} \) and 16 \( \mu\text{m} \), respectively. However, the droplet concentrations for coughing and speaking were 2.4 to 5.2 cm\(^{-3}\) for each cough and 0.004–0.223 cm\(^{-3}\) for speech [6].

Gupta et al (2009) focused on studying the flow dynamics of human cough which is believed to help prevent the transmission of infectious airborne diseases. Computational fluid dynamics simulations were carried out to predict the passage of the virus considering flow rate and direction, mouth or nose opening area and temperature. Two types of flow measurements, single and sequential cough were examined among male and female test subjects to interpret the accurate measurements. It was important to have high frequency measurements (100Hz or higher) as the peak velocity time was in the order of milliseconds. Cough flow rates were measured using a
spirometer. Single cough resulted in less inhalation volume while sequential cough resembled a combination of single cough. The results varied between male and female subjects and concluded that physical parameters such as height and weight were important consideration to be made. The mouth opening area, flow direction and mean angles of a cough were obtained from flow visualization. The reference cough velocity was taken to be in a range of 6 to 22 m/s (Zhu et al). The gamma-probability-distribution function for each cough was defined with the help of non-linear least square curve fitting analysis in MATLAB. It was concluded from this paper that height, weight, gender of a person and ambient temperature which was found to be an influential parameter were major inputs for the CFD simulations and anticipated to use mean mouth area during cough [16].

Generation of aerosols in a medical setting can also occur from numerous medical procedures, which are known as aerosol generating procedures. These procedures include tracheal intubation, non-invasive ventilation, tracheotomy, manual ventilation before intubation, bag valve mask ventilation, and nebulizer therapies [4]. For example, Khai tran et al (2012) [4], study to evaluate the risk of health care workers being infected by treating patients undergoing aerosol generating procedures compared to not undergoing these procedures as mentioned above. The authors further characterized the study of these methods into the period of evaluation, the population, types of laboratory tests to confirm the diseases, and assessment of training and protection equipment use. Health care workers were confined to various laboratory tests and results indicated tracheal intubation posed a high risk in transmitting the aerosol particles among other activities. This could be considered as a reason for health care workers being in proximity and treating the patients. There exist several limitations in this paper by lack of quality in design, imprecision and was not generalizable. The authors concluded by stressing on how important it is
to provide funding for high quality research in health care. Inhalation and deposition of virus-containing droplets or aerosol particles is the final stage of transmission.

1.6 Previous Examples of Simulator Designs

Previous research on models (or mannequins) which simulate various respiratory flows will now be reviewed. These models have been used to study air flow patterns generated by respiratory activity. Similar models that can generate aerosols have been designed to study the transmission and flow pattern of particles generated by respiratory activity.

Lindsley et al (2013) explores the mechanism of generating cough aerosol particles by designing a cough aerosol simulator in order to study how these droplets play a major role in transmitting infectious diseases. The design and working of the simulator are a first of its kind and function on steel bellows run by a computer driven linear motor. The bellows were moved upwards along the shaft and then down pumped with test aerosol with the help of a vacuum scavenger, prompting cough. Experiments were conducted by synchronizing the cycles of the coughing and breathing simulators such that at every inhalation cycle, cough is initiated. Measurements of expiratory volume and flow rate were noted using an ultrasonic spirometer. Further, droplet size distributions were measured using a spray droplet size analyzer [8].

Kim and Chung (2015) establish a very precise model of the whole of the human airway consisting of nasal cavities, larynx, trachea, and two generations of bronchi to study the flow characteristics of nasal breathing. The airway model was designed with the help of a high-resolution CT scan and rapid prototyping machine. It is said to be a functionally correct model when compared with the rest of the models around the world. The main body of the airway model consists of a cylindrical pump and cam-controlled piston. The respiratory cycle was divided into 4 phases while the measurements taken by particle image velocimetry (PIV) were divided into 7.
The PIV measurements of the working fluid (a mixture of water and glycerin) flowing in the dynamically scaled model were taken at the nasal cavity, larynx and trachea, and bronchiole. The experiments and the results concluded that a thorough designed model can obtain a realistic flow condition which could help doctors and other agents understand and treat disease efficiently. However, these authors did not study the flow expelled from the mouth or nose, focusing instead on internal flows in the respiratory system [12].

Feng et al (2015) study a breathing thermal mannequin to study the flow characteristics of the inhaled and exhaled air. A phase averaged method and 2D PIV was used to measure and analyze the flow pattern and dynamics. PIV in this experiment could study turbulent characteristics and boundary conditions unlike the others. The mean velocity, turbulence and vorticity from this study would be used for data processing and making a turbulent mode to validate it by CFD analysis. The mannequin breathed out 0.45 L of air in one exhalation with a mouth diameter of 12mm. Tests were also conducted to understand the thermal buoyancy in heated and isothermal conditions. The velocity profile was bell shaped during exhalation and flat during inhalation. The turbulent intensity is weak in heated conditions and strong in isothermal conditions [17].

Zhang et al (2017) investigated the transmission of aerosols by designing a cough simulator expelling transient flow making it more realistic to human respiration. A pressurized gas tank, nebulizer, solenoid valves were part of the model. Coarse and fine droplet size distributions were examined through laser diffraction method. The cough simulator could generate an aerosol of 800 µm. The study reveals that evaporation could be negligible for larger droplets. The volume capacity ranges 0.25 to 3.83 L cough, the velocity of the cough ranges 5 to 10 m/s and the velocity of coarse particles ranges 1.9 to 4.2m/s [23].
Verma et al (2020) conducted experiments to test the efficiency of the face coverings to minimize the droplet dispersal using a mannequin simulator to simulate coughs and sneezes. Visualization experiments are conducted on three types of face masks with the help of a pump and fog/smoke machine to generate particles. It was observed that the droplets not only travelled farther than 6 ft but were also suspended in mid-air for a longer period of time than expected. There were significant leakages observed around the edges and through the cloth and cone-style masks which are currently being used by significant numbers of people. Hence masks made with multiple layers, the correct material, filters, social distancing, and other measures are important to be understood by the health care workers, manufacturers, and the consumers [3].

Almost all the models so far have been designed to examine singular breathing events such as cough or sneeze or by speaking. Further, the one study which did use a mannequin model to study breathing did not examine aerosol dispersion (Feng et al 2015). However, breathing is a continuous activity and, though the number and concentration of particles may be lower, the longer time over which this activity occurs may lead to a greater transmission risk than relatively short duration respiratory events. This study involves generating aerosols by designing a breathing simulator and measuring the particle size, concentration, and distance with the help of TSI Optical particle sizer and flow visualization with the help of laser light.
Chapter 2 Design of Breathing Simulator

2.1 Overview

This thesis involves the study of generation and transmission of aerosol particles by designing a tidal breathing mannequin simulator. It is important to understand the viability of the particles carrying pathogens after their emission and how the virus transmits in common spaces when breathing and speaking. The model is used to conduct physical experiments of dispersion of aerosols with and without the use of face masks. By testing and understanding the effectiveness of face coverings, the spread of virus can be controlled. The aerosol size distributions and concentrations were measured with TSI Optical particle sizer and will be discussed in the next chapter.

Figure 1 CAD Model of Human Breathing Simulator
Figure 1 shows a CAD model of the human breathing simulator used in this study. The system was designed and built-in collaboration with Senior Capstone Teams in Fall 2020 and Spring 2021. The system is controlled by the Arduino Uno and control board to operate the linear actuator which expands and compresses the two bellows to produce human-like breathing. Two solenoid valves are used for the intake and release of air through mouth and nose of the mannequin. The air is drawn into the system by the inhalation solenoid valve when the bellows expand and pushed out into the vinyl tubing during compression. This air flow passes through the heat exchanger via the exhalation solenoid valve to maintain the humidity and temperature of realistic human breath. The humidified and heated breath is pushed through the spirometer to measure the flow rate. The breath may then be expelled through either the mouth or nose; the flow direction can be controlled by a Y-splitter based on the requirements. The temperature of air in the heat exchanger and at the mouth are measured by sensors.

2.2 Design Requirements

The main criteria for designing the model was to replicate human tidal breathing pattern. This includes a two-way breathing system where inhalation and exhalation take place at two spatially separated points. The purpose of this spatial separation is so that aerosol particles may be injected into the flow system to better study the exhalation. Further, the exhaled air should have a biologically realistic temperature and relative humidity. This requirement is because moist human breath is warm and therefore buoyant, which has been shown to affect the respiratory airflow. The model is such that the aerosols can be treated or injected with a generator and the concentration at the output air flow can be measured. The air flow can also be directed either through the mouth or nose so that the effect of breathing through either may be studied. The flow profile (i.e flow rate or flow speed over time) also should be realistic and measurable.
2.3 Construction

Figure 2 Components of the Breathing Mannequin Simulator

Figure 2 consists of the design and parts of the breathing simulator. These parts include the Arduino Uno in conjunction with the control board, the linear actuator that compresses and retracts, the pairs of bellows, two solenoid valves, a humidifier and heating pad, and a spirometer. Tubing of 3/8” inch inner diameter and 1/2” outer diameter is used from the bellows which splits into a T-splitter and connects to an exhalation solenoid valve and to the humidifier. Each component will now be described in detail.

2.3.1 Spirometer

A spirometer is a medical device used to assess lung function by measuring the inhalation and exhalation of air volume of a full breath. This device helps in examining the health of the respiratory system and is also effective in diagnosing respiratory diseases. A FSP20 Spirometer purchased by Facelake is small in volume, consumes less power, has a display screen and is portable. It comes with a mouthpiece and turbine that are detachable. The measurements are taken
by breathing out all the air into the mouthpiece as fast as possible and the screen will display the measured parameters like Forced vital capacity (FVC), Forced volume in ONE sec (FEV1), and Peak expiratory flow (PEF). This spirometer works by using the air exhaled into the mouthpiece to rotate turbine blades. These blades are judged by the infrared emission and reception tube inside the device and transform the light signal received from the blade motion to measurement parameters displayed on the screen. The spirometer may measure a volume range of 0 – 10L, flow rate of 0–16 L/s, has volume accuracy ±3% or 0.5 L/s, and flow rate accuracy of ±5% or 0.2 L/s. The turbine must be aligned with respect to the turbine hole and locked by rotating it clockwise and the mouthpiece must be inserted into the turbine port. While measuring, the screen displays testing and after a few seconds the measured parameters will be displayed on the screen and when choosing the downward button will display the flow rate volume charts.

2.3.2  Linear Actuators

The 12V High Speed 4in Actuator of make PA-14P-4-35, is used to compress and retract two bellows for 3.5 in and has a stroke of 4 in and speed of 2 in/s, generating sufficient air circulation within the system. The speed and motion of the linear actuator can be controlled by changing the code and entering a new range of values 185 - 255 for PWM function (float amplitude).

2.3.3  Bellows

The bellows are used to pull in fresh air through the inhalation solenoid valve during retractions and provide a strong blast of air into the system with the help of linear actuator during compressions. They are flexible elements that absorb movements into the pipe system. They are to represent a pair of human lungs that can generate the amount of air a human can in one breath, which is close to 0.5 L/s. Initially a single bellow was used to generate the air flow, but this could
not generate any readings in the flow meter and an additional bellow was added to the system to achieve sufficient flow rate. Plunger bellows were used to pump in air in the system and these bellows are horizontally rigid under compression, inexpensive, and easily modified. One end of the bellow is attached to the linear actuator and the other to the tubing that transports air throughout the system.

2.3.4 Solenoids/ Flow System

Solenoid valves are used for controlling the flow of a fluid in the mannequin system. Two 2-way 1/2" Stainless Steel Electric Solenoid Valves of 12V DC model - USS2-00069 ordered on Amazon are used in the model, one for the inhalation of air pumping into the system and the other for exhalation flowing out through the mouth and nose of the mannequin. Since a higher rate of air flow was required in the system, valves of ½” npt were chosen and can be controlled and operated with ease. The solenoid can be opened or closed by energizing or de-energizing it. When powered on, magnetic field enables the plunger to rise, and the unsealed valve allows the fluid to flow through the system and when the power is cut off, seals the valve preventing the flow.

The first valve i.e the inhalation valve pumps in air into the 3/8”in tubing and into the bellows. The second valve is positioned near the humidifier/ heat exchanger and the resulting outcome is that of a heated and humidified air flow.

2.3.5 Humidifier/Heating Reservoir

It is important to incorporate a heat exchanger in the system to generate human breathing with biologically realistic temperature, humidity, and flow rate. The heat exchanger consists of an inductive heating pad which heats the reservoir filled with water. Due to the heated water, humidity builds up in the reservoir and the air introduced into it, gets heated and humidified. This heated and humidified air is pushed into the spirometer to measure the flow rate. The temperature at the
reservoir and that of the exhaled air at the mouth of mannequin is monitored by temperature sensors. There are 3 inlets in the reservoir, for the air flow, sensor and another for particle generator. This generated sufficient humidity of 96.1 % which falls into the range of human breathing of 95% ± 5%. Safety precautions must be used when operating the heating pad because of the high temperatures involved.

2.3.6 Mannequin Head

The mannequin head was designed using CAD such that the flow could travel through the mouth as well as the nostrils. The mouth was modeled as a round hole with a diameter of ½ in. However, in the experiments described below, a tube with an inner diameter of 3/8 in and outer diameter of ½ in was threaded into the mouth up to the lips. The airflow was thus emitted from a circular hole with diameter of 0.9525 cm. Nasal cavities was designed similarly to human anatomy, which have a more complex morphology than a simple circular tube. The 3/8” vinyl pipe was threaded into the nose similar to the mouth and the flow through the cavity branches out to a pair of nostrils, which had a somewhat elliptical cross-section with a long axis of 0.7 cm (total diameter of both the nostrils 1.4 cm). The expelled air jet then flows through the nostrils. The direction of flow can be controlled by a Y splitter.

2.4 Testing

Prior to testing the Covid-19 mannequin, the tubing running through the entire system as shown in Figure 2 was connected and sealed to minimize flow loss. The humidifying chamber was filled with water and the heating pad was turned on and set to 150°F to heat the water. The temperature sensors were placed at the reservoir and mouth to measure the temperature of the incoming and outgoing air flow. Once the temperature at the reservoir reached ±32°C in 20 to 30 minutes, the system was plugged in. The spirometer was turned on to measure the flow rate
readings of the air passing through the system. The temperature of the exhaled air was noted. The connections between the solenoid valves and the Arduino Uno were re-checked during a few tests run, to ensure no obstructions existed for the fluid flow.

Figure 3 Time Series of Flowrate on Spirometer

Figure 3 shows a sample time series of the flow rate measured using the spirometer. The peak flow rate is 0.74 L/s, which is similar to measurements taken from the respiratory flow of a human during testing [4]. The shape of the curve is like the shape of the curve in the study conducted by Kim and Chung (2015) or a natural breathing curve.
Chapter 3 Flow Visualization

3.1 Introduction

Experiments were conducted in indoors, in a classroom, to analyze the mouth and nasal flow patterns of the breathing mannequin simulator with and without face coverings. A fog machine was used to generate tracer particles to analyze respiratory jets using a mixture of humidified air. The smoke was pumped into the system by connecting it to inhalation solenoid valve and expelled through the hollow nasal cavity and mouth from the mannequin head via vinyl tubing.

A light illuminating source known as the Green Lantern was used to visualize the flow of the expelled particles through the mannequin head. A high-speed camera (Nikon) with a resolution of 1920 pixels × 1080 pixels and frame rate of 60/sec was set up to capture the flow pattern.

3.2 Experimental Set Up

The experimental set up consists of the mannequin breathing simulator, a light source known as the Green Lantern, a high-speed camera, fog generator, and the mannequin head placed in front of a black backdrop to visualize the flow pattern, as shown in Figure 4. The Green Lantern is an LED-based light source which provides an approximately 1 cm thick sheet of light. This device is 74.3 cm long and emits 51.7 lm, thus providing plenty of light for visualization purposes. The LED light source was placed at 70 cm from the mannequin on the left (out of the frame) aiming at the vertical center line of the head that expels the aerosol particles as shown in Figures
5 and 6. As the smoke was generated from the machine and pumped into the system via inlet solenoid, the simulator is powered to expel the flow passing through it. A camera was set up to observe the flow that can be seen with the help of the green lantern, as shown in Figure 6. The flow pattern is observed for the following cases: flow through the mouth with and without mask and flow through the nose with and without mask. The significance of wearing a mask on expelled respiratory jet can be observed in flow visualization. The purpose of wearing a mask is to obstruct the flow and reduce the spread of any infectious airborne particles. Although there are many kinds of masks available in the market, a fabric mask is used in the following experiments and the flow pattern can be observed.

The flow rate for each case is measured by the spirometer at the initial phase of each experiment. Figure 5 shows an example of the fog emitted as a jet from the mannequin mouth as it is illuminated by the Green Lantern.

![Figure 4 Experimental Setup for Flow Visualization](image)
3.3 Results

The following sections are the results of the four cases mentioned previously and the duration of breath considered is 3.3 s long. Since the frame rate was 60/s, there were a total of 190 frames extracted for each case.
3.3.1 Case 1 Mouth with No Mask

The series of images shown in Figure 7 is the flow pattern observed for the case of gas being expelled from the mouth without a mask. The images are chosen at 12 different time intervals that are mentioned in the Figure 7 with a peak flow rate of 0.61 L/s. There is no flow observed at \( t = 0 \) s. The first image considered at 0.0173 s (17.3 ms) shows a jet expelled a horizontal distance of 2.29 cm from the mouth. The flow continues to surge forward to 19.38 cm at 0.33 s. At \( t = 0.764 \)
s, the flow developed as a turbulent jet and has traveled a total distance of 32.85 cm. At t = 1.66 s, the flow extends beyond the field of view (i.e., > 50 cm). By t = 2.51 s, the exhalation has largely stopped. The jet velocity then gradually decreases, and the jet ends with a cloud of particles which are swirling around as a turbulent flow far from the mouth. The flow ends at t = 3.21 s and comes to a complete halt at t = 3.3 s before the next cycle begins.

![Graph showing distance covered vs time for the front of the expelled jet in all four cases.](image)

**Figure 8 Comparison of Distance Covered by Expiratory Jet in all Four Cases Against Time**

Figure 8 shows the horizontal distance travelled vs time for the front of the expelled jet in the 4 cases. The blue curve is the first case which travels the farthest distance (i.e., > 40 cm) compared to the 3 other cases at time > 1s. The maximum speed of the jet is approximately 1.34 m/s at t = 0.069 s, which gives a Reynolds number of 775, where the tube diameter in the mouth is taken as the characteristic inner diameter of 0.9525 cm (3/8 in). This Reynolds number places the exhalation jet in the laminar regime.
3.3.2 Case 2 Mouth with Mask

Figure 9 Flow Pattern of Mouth with Mask

Figure 9 represents the image sequence of dispersal pattern of expelled respiratory jet when the mouth is covered with a fabric mask. The peak flow rate of this cycle is 0.72 L/s. Compared to the flow pattern of mouth with no mask, the jet in this case is obstructed significantly due to the
mask. The leakage of flow particles through the mask can be observed initially at \( t = 0.0173 \) s and continues to build up throughout the sequence. The jet at \( t = 0.503 \) s has traveled a horizontal distance of 4.03 cm and can be seen rising upwards slightly. The upward flow of the jet may be due to the buoyant warm air flowing out the mouth due to density difference and a temperature of 35° [22]. The jet rolls up to form a mushroom-shaped vortex between the time frame of 1.094 s and 1.66 s. The formation of this vortex is possibly due to the air exiting the tube inside the mouth and a faster decay of the shape due to thermal plume and thermal buoyancy [17]. Even though the leakage of the flow through the mask is minimal, particles escape around the edges of the mask as seen at 1.98 s and 2.51 s. Most of this flow escapes from the gaps at the nose and chin area of the mannequin head. The horizontal distance traveled by the jet is shown in Figure 9. In comparison to the case without a mask, the travel distance is significantly decreased. The maximum flow speed occurs at \( t = 0.191 \) s, which gives a Reynolds number of 104 for the flow going through the mask, which is significantly less than the previous case. In general, the emitted flow stays quite close to the mannequin head.

3.3.3 Case 3 Nose with No Mask

Figure 10 represents the flow pattern expelled through the nose without a mask. The flow through the nose takes place by blocking the flow through the mouth by using a Y splitter. The air is expelled at a peak flow rate of 0.6 L/s at 36.4°C. Due to the nasal cavity, the jet is observed to flow in a downward direction in two jets, one forming from each nostril. At initial stage of 0.0173 s, the emitted flow, which can be seen pointing downwards, has traveled a horizontal distance of 1.83 cm. The area of the two nostrils is greater than the diameter of the tube in the mouth. The resulting jets therefore have different characteristics, with the jet through the nostrils being wider and spreading more rapidly as it entrains the flow around it. The jet builds up momentum and reaches the maximum horizontal distance of 20 cm at 1.40 s before it starts to dissipate gradually.
The flow starts to move further downward until 1.98 s and slows down and halts the cycle at 3.21 s. The rightward progression of the fog during this time may be because of stray air currents in the room.

Figure 10 shows that the horizontal travel distance (19.5 cm at t = 1.985 s) is much less than the case of the jet emitted through the mouth without a mask but is greater than the case of the jet emitted through the mouth with a mask. Overall, this flow has a lower horizontal speed as well, with the maximum horizontal speed of 0.35 m/s occurring at t = 0.55 s. This speed is obviously lower because the jet is directed downwards at an angle. (The maximum speed of the jet along the direction of the flow is 0.82 m/s.) It also is lower because the jet is emitted through a larger aperture (i.e., through two nostrils). The Reynolds number based on the horizontal velocity component at this maximum speed is 295, but the Reynolds number based on the maximum velocity along the trajectory of the emitted smoke is somewhat greater (Re = 692).

3.3.4 Case 4 Nose with Mask

Figure 11 represents the flow pattern of expelled air through the nose with a mask. Unlike the flow through nose without the mask, the expelled jet is reduced significantly. Some fog from the previous cycle is visible at t = 0.0173 s at the ridge of the nose. The emitted fog slowly moves through the mask from t = 0.33 s to 1.094 s with a maximum horizontal travel distance of 7.19 cm. However, most of the tracer particles escape through the edges of the mask and flow towards the frame of view. This could probably be taken as the distance the jet might have travelled which is not the case. The peak flow rate of the jet in this cycle is 0.77 L/s. There is a significant leakage of the flow at the top edge between the nose and the mask as shown in Figure 12, on either side of the mannequin head and at the bottom of the face covering at 1.95 s to 2.72 s. The maximum speed in this case was 0.25 m/s at t = 0.72 s, with a Reynolds number of 211.
The jet travels only 3.5 cm compared to 9 cm travelled in the previous case at the same time (formation of vortex in the flow of mouth mask at same time). The maximum speed is attained later than the case for nose with no mask, probably due to the obstruction of flow by the face mask.
Figure 11 Flow Pattern of Nose with Mask
3.4 Analysis

One of the measures considered in this study is the horizontal distance travelled by the expiratory jet in each case. This is because it is the distance travelled by the contaminated aerosols/flow that could reach and infect to other people that matters in the transmission. Horizontal travel distance of particles was observed to be maximum for the case of the mouth with no mask at a distance > 60 cm beyond the frame of view. The horizontal travel distance was less for the case of nose with no mask, largely because the expiratory jet was directed downwards. The mask used here significantly reduced the horizontal reach of the expiratory jet for both the mouth and nose breathing cases. Even though air leaked around the edges of the mask in both cases, it was effective in keeping the expired flow (and the aerosol particles within that flow) closer to the body of the person, which could potentially help to slow transmission of airborne diseases. Considering the speed and Reynolds number of the flow is another important factor that helped in determining the flow dynamics.

3.5 Discussion

Air flow patterns look qualitatively similar to those found by Feng et al (2015) in a breathing thermal mannequin. However, their results showed more upward rise in the trajectory owing to the buoyancy of the heated air. This may be because of the inconsistent pumping of smoke into the system or the air flow not being warm enough during that cycle. Also, the maximum exhalation speed attained in this experiment comparing all the 4 cases was 1.3 m/s through the mouth at 35°C compared to a human exhalation speed of up to 4 m/s as measured by Feng et al (2015) at 34°C. Feng et al experiments were limited to inhalation and exhalation through mouth whereas this study conducted experiments for exhalation for 2 cases nose, and mouth. Similarly, results obtained by a study conducted by Gupta et al (2009) expressed the exhaled flow velocity
over time to be a sinusoidal function. The speeds of exhaled air were expected to be lower due to speed of the linear actuator which curbs the flow rate of air in the system. Turbulence intensity was observed in Feng et al due to a sampling frequency of 1000Hz which is higher compared to this study and turbulence does not apply much for small sampling frequencies.

However, study conducted by Kim et al (2015) resulted in a flow rate ranging 0 to 0.4 L/s (from model) which is less than the achieved 0.83 L/s close to a human breathing with a flow rate of 1 L/s (from human). The Reynolds number achieved in these experiments were in range with Kim et al (2015) 0 – 800. It is evident from the sequential images that the flow pattern and speeds are obstructed due to the face coverings. The maximum peak flow rate comparing all the 4 cases was at 0.77L/s for nose with mask followed by 0.72 L/s for mouth with mask. The flow pattern is seen to be transient in case 1 and turbulent in case 2 due to the vortex formation.

The peak flow rates achieved in this experiment (i.e., 0.86 L/s at 2cm) are close to the range of an actual breath for the case nose with mask as it states that the average human flow rate of air is about 1.0 L/s [12]. The pressure rise in the system was comparatively less than expected due to the linear actuator having a maximum speed of 2 in/s. Improving the speed of the actuator will give access to an improved control of the breathing frequency. The expelled air jet through the narrow passage of the nostrils restricted a smooth and continuous flow. An improved version of a realistic nasal and mouth cavity could enhance the flow pattern to be continuous. The flow visualization is studied for exhaled air through mouth and nose unlike other studied referred to.
Chapter 4 Particle Sizing Measurements

Physical experiments of the dispersion of aerosols emitted from the breathing mannequin simulator with and without protective face mask in an enclosed classroom is utilized to compare the results with that of actual human breath. Aerosol size distributions and concentrations are measured by a TSI optical particle sizer (OPS), and the resulting aerosol particle ejection rate from the mannequin will serve as input for further analysis. Droplet dispersion throughout the classroom by the model will be compared with the physical measurements. This chapter focuses on just the particle concentration and its size range as they play the most important role in analyzing spatial spread/transmission.

4.1 Experimental Setup

![Experimental Setup of Mannequin without Face Mask](image)

Figure 12 Experimental Setup of Mannequin without Face Mask
Figure 13 Experimental Setup of Mannequin with Face Mask

Figure 12 and Figure 13 represents the experimental setup for measuring the particle concentration and size with and without a face mask. The experimental setup for measuring the particle sizes and concentration is as shown in Figure 12 and Figure 13. The OPS could either be mounted on an adjustable tabletop or placed farther away from the mannequin head with a conductive tubing than can be connected to the inlet nozzle of the OPS. This tube could be pointed towards the expiratory jet with the help of clamps. Aerosol instrument manager software was installed in a computer to extract the data measured by the OPS.

When the breathing simulator was powered, the software in the system was manually operated to start recording the measurements for 60 seconds. During this time, the fog generator that was used for the flow visualization also was operated to produce particles for the current measurements. Measurements at five different horizontal distances from the mannequin mouth were acquired for the four conditions previously examined in the flow visualization chapter. These
distances are mentioned in Table 1. The peak flowrate also was recorded for each experiment, and Table 1 gives these results.

<table>
<thead>
<tr>
<th>Distance (cm)</th>
<th>Mouth No-Mask (L/s)</th>
<th>Mouth With-Mask (L/s)</th>
<th>Nose With No-Mask (L/s)</th>
<th>Nose With-Mask (L/s)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.77</td>
<td>0.57</td>
<td>0.76</td>
<td>0.67</td>
</tr>
<tr>
<td>2</td>
<td>0.63</td>
<td>0.58</td>
<td>0.78</td>
<td>0.86</td>
</tr>
<tr>
<td>5</td>
<td>0.61</td>
<td>0.72</td>
<td>0.61</td>
<td>0.62</td>
</tr>
<tr>
<td>10</td>
<td>0.64</td>
<td>0.71</td>
<td>0.73</td>
<td>0.66</td>
</tr>
<tr>
<td>50</td>
<td>0.62</td>
<td>0.74</td>
<td>0.68</td>
<td>0.82</td>
</tr>
</tbody>
</table>

The ambient level of particle concentration in the room also was measured by placing the OPS at a distance greater than 50 cm from the mannequin after all experiments were conducted. A description of OPS and its working is explained in the following section.

4.2 Optical Particle Sizer

TSI optical particle sizer (OPS) spectrometer Model 3330, shown in Figure 15, is used to measure aerosol size distributions and concentrations. It comes with power cables, battery, conductive tubing, zero filter, aerosols instrument manager software CD-ROM, a user manual, etc. to run the experiments.

Figure 14 is the isometric view of the TSI Optical particle sizer. It works on the principle of single particle counting, by detecting particles with the help of a laser and a detector. The flow rate measured is 1L/min. The pump in the system circulates the sheath flow and helps in focusing the laser light on the particles and prevents their contamination. The particles pass through a laser beam and the light scattered by the particles is focused on the mirror. This is sensed by the photo detector and the rest of the process is performed by electronics. The device can count 50% of 0.3 \( \mu m \) particles and 10 \( \mu m \) is the maximum size of the particle that can be measured. The particles
processed through the device are trapped in a filter which can be used for further chemical analysis.

The instrument can measure the count of total number of particles, total number concentration in particles per cubic centimeter (#/cm³), total particle mass concentration in micrograms per cubic meter (µg/m³), charts representing concentrations with respect to particle sizes, counts (dC), Number concentration (dN), Mass concentration (dM), linear and logarithmic graph. The data is measured in table and graph format as shown in the Figure 15 which is a reading of the case mouth with no mask at 5cm from the mannequin. The units of measure can be changed as required and a sequence of samples can be measured. The data can be stored by connecting OPS to a computer via USB cables.

4.3 Data Processing

An example of how data is processed from the OPS is shown in Figure 16. It represents the data collected for the case of flow through the nose with the mask at 50 cm. The table to the right of figure 17 shows the data of number concentration in units of particles/cm³ and the particle size bins in units of µm, which are the key measures required for further analysis. The OPS presents the data in 15 size bins.
These bins were grouped into 3 larger bins (representing small, medium, and large particles), each comprising 5 of the original bins. The bin with smaller particles range 0.337μm to 1.007μm in
diameter, bin with medium range particles 1.254 µm to 3.752 µm, and bin consisting of larger
droplets range from 4.672 µm to 9.016 µm. The five particle concentrations in each of these bins
were averaged to produce an average particle concentration for the small, medium, and large
particles. These average particle concentrations could then be compared with all the 4 cases.

4.3.1 Average Concentration vs Distance for Mouth and Nose, with and without Face

Coverings

Figure 17 Average Concentration vs Distance for Mouth with No Mask. Ambient Average Concentration of Particles in Bins 1, 2 and 3 are 6.7, 9.2 and 0.3 particles/cm³

Figure 17 shows the average particle concentration in three bins versus distance for the
mouth with no mask. The particle concentration levels (which are almost above the ambient level)
decrease with increasing particle size for all distances from the mannequin. Maximum average
concentrations for the smallest particles (Bin 1) are about 350 particles/cm³, decrease to 130
particles/cm³ for the medium particles (Bin 2), and decrease substantially to about 2 particles/cm³
for the large particles (Bin 3). The concentration levels for particles in Bins 1 and 2 (0 – 3.7µm)
are somewhat similar, with a difference of only approximately 100 particles/cm$^3$. For all size classes, there is an overall decrease in concentration with increasing distance from the mannequin, though it should be noted that the concentration increases slightly for the small particles (Bin 1) as distance increases from 1 to 2 cm. The particle concentrations for all size bins are somewhat greater than (Bins 1 and 2) or reaches the ambient levels at the further distance measured.

![Figure 18 Average Concentration vs Distance for Mouth with Mask.](image)

Figure 18 Average Concentration vs Distance for Mouth with Mask. Ambient Average Concentration of Particles in Bins 1, 2 and 3 are 6.7, 9.2, and 0.3 particles/cm$^3$.

Figure 18 shows the average particle concentration in three bins versus distance for the mouth with a mask. Again, the particle concentration levels generally decrease with increasing particle size, though the particle concentrations are similar for Bins 1 and 2 initially at 5 cm away from the mannequin. Overall, the particle concentrations are lower for the case with the mask as compared to the case without the mask. For the small particles, average particle concentrations decreased from about 350 to below 100 particles/cm$^3$. Medium particles experienced a similar decrease. The large particles (Bin 3) experienced the greatest decrease, going down two orders of
magnitude at the closest distance to the mannequin (and reaching below ambient levels near the mouth). These results show that the mask is particularly effective at blocking large particles. Now considering how particle concentrations vary with distance, Figure 18 shows an increase of concentration levels in the curves for all particle sizes transitioning between 1cm to 10 cm. Maximum concentration are observed at 50 cm. The trend among the increasing concentrations is due to smoke escaping around the mask away from the face rather than through the mask. The smoke escaped around the nose region and downward and around towards the measurement area after a certain time which is why a peak in concentration is observed far from the mannequin. This observation can be backed by Figure 9 (flow pattern of mouth with mask). Comparing both the cases of mouth with and without mask, the far away concentrations are similar whereas overall concentration of Figure 18 is less than Figure 17.

![Figure 19](image.png)

Figure 19 Average Concentration vs Distance for Nose No Mask. Ambient Average Concentrations of Particles in Bins 1, 2, and 3 are 6.7, 9.2, and 0.3 particles/cm³.
Figure 19 shows the average particle concentration in three bins versus distance for the nose with no mask. Small and medium particles in the first 2 bins (ranging 0 µm to 4 µm) are an order of magnitude lower than the results from the case of the mouth with or without the mask. Indeed, the concentrations are close to the ambient range. The large particle concentrations are somewhat similar to those observed previously. These overall low concentrations are probably due to the direction of expiratory jet in downward direction expelled through the nostrils and not being directed horizontally compared to other cases. No particular trends are seen in how these concentrations change with distance from the mannequin, again likely because the jet with the particles is directed downwards.

![Figure 19: Average Concentration vs Distance for Nose with No Mask.](image)

Ambient Average Concentrations of Particles in Bins 1, 2, and 3 are 6.7, 9.2, and 0.3 particles/cm³

Figure 20 shows the average particle concentration in three bins versus distance for the nose with a mask. The smaller particles (Bin 1 and Bin 2) have greater concentrations than were observed in the case with the nose without a mask. Indeed, these concentrations are similar to those
found for the case of the mouth with a mask. Considering that the flow patterns are similar for the mouth and nose with the mask, it is perhaps not surprising that the particle concentrations would also be similar. The mask effectively decreases the particle concentration close to the mannequin face. The large particle concentrations also are similar to those previously observed. Particle concentrations for Bins 1 and 2 decreases with increasing distance from the mannequin but seem to increase with increasing distance for Bin 3. It is assumed that the smaller particles are close to the face and were being sampled there, resulting in decreased concentration compared to larger particulate concentration.

4.3.2 Comparison Graphs for Average Concentration vs Distance for Mouth and Nose, with and without Face Coverings

![Comparison Flow of Four Cases of Particle Size Ranging 0.337 µm - 1.007 µm](image)

Figure 21 shows a comparison of the average particle concentration for small particles (Bin 1) for all four tested cases. Similar plots for the medium particles (Bin 2) and the large particles (Bin 3) are shown in Figures 22 and 23, respectively. These plots provide a useful way to compare
results among similarly sized particles for the four different cases. This is important because different particles behave differently. For example, large particles (Bin 3) are more likely to be caught by the mask or to fall to the ground under the effects of gravity. The decrease in concentration of small particles is evident in cases without a mask near the face region. The concentration of large particles is observed to increase and fall away from the face as shown in Figure 21 (blue and grey curves).

![Graph showing particle concentration vs. distance](image.png)

**Figure 22 Comparison Flow of Four Cases of Particle Size Ranging 1.254 µm - 3.752 µm**

Figure 22 shows a comparison of the average particle concentration for medium range particle (1.254 µm - 3.752 µm). The curves for the first 3 cases in Figure 22 have a similar trend with respect to concentration compared to the case nose with mask. There is also an increase in concentration for cases without mask for mouth and nose at around 10 cm. No difference is observed in the trend at farther distances.

Figure 23 shows a comparison of the average particle concentration for large range particles (4.672 µm - 9.016 µm). It can be observed that the concentration of larger particles
decreases over distance. This is probably due to the mask catching larger particles leading to a lower concentration which is around 1 particle/cm³.

Figure 23 Comparison Flow of Four Cases for Particle Size Ranging 4.672 µm -9.016 µm

4.4 Discussion

All the graphs plotted in the previous sections compare the concentration of particles and particle size range (i.e small and large particles) with respect to the distance travelled. Further, the results from flow visualization and the OPS readings were compared and observed. All the calculations were made at a constant temperature of 35°C.

The study conducted by Feng et al (2015) is similar to this study but the peak velocity measured here is somewhat lower. Feng et al (2015) measured a peak flow speed of approximately 4 m/s using PIV whereas the peak flow speed measured here, using the flow visualization images, was 1.3 m/s. This difference may be attributed to differences in measurement technique, differences in expiratory flow rate, and differences in mouth size. Since the mouth diameter in the
current study (9.5 mm) was smaller than mouth diameter used by Feng et al (12mm; 2015), the observed difference in peak speed is likely due to the lower flow rates used in the current study. It was very important to conduct experiments considering the temperature as one of the conditions.

Few studies have examined the dispersal of droplets by breathing, so the results from the current study will be compared to results from studies that examined coughing. Even though the study by Gupta et al (2009) examined the exhaled flow, experiments based on cough were considered in that study. Similar experiments on coughing were conducted by Verma et al (2020), where the flow travelled an average distance of 8 ft (without a face mask) and 8 in wearing a commercial mask. When compared to the results obtained in the current study for breathing, the maximum distance the expelled air jet travelled with a mask was 20 cm, which was similar to the distance found by Verma et al (2020). However, without a face mask, the maximum distance traveled in the current study was greater than 60 cm. The true maximum distance extended beyond the field of view but likely did not reach 8 ft. The velocity of the air and the carried airborne particles are also much lower for breathing than for coughing. The maximum speed measured here was 1.3 m/s, but Zhang et al (2017) found the velocity of the cough ranges from 5 to 10 m/s and the velocity of coarse particles ranges 1.9 into 4.2 m/s. For the breathing simulator studied here, the results are somewhat lower than the reference range, as expected.
Chapter 5 Summary and Future Directions

The tidal breathing mannequin simulator was designed and developed to study the flow pattern, aerosol concentrations, and size distributions of exhaled air jets through the nose and mouth. The breathing model was designed to attain an exhaled breath with realistic temperature, relative humidity, and flow profile comparable to a human. Even though the studies conducted previously analyze a cough or sneeze, they happen to be single breathing events with not much work done on breathing which happens to be the easiest and simplest way to transmit airborne diseases. The difference with this study is that the novel Covid-19 breathing simulator can expel air through mouth or nose (via Y splitter), a design feature that lacks in other models. Flow visualization and aerosol size and concentration measurements were conducted with the help of laser light and TSI Optical particle sizer.

An important factor considered while conducting these experiments was to study the dispersion of aerosols with and without face masks in all the cases. Flow visualization experiments were performed with the help of Green Lantern and Nikon camera to understand how the flow through nose and mouth takes shape with respect to distance and time. It is important to consider the horizontal distance travelled by the air jet as this flow carrying contaminated aerosols could reach and infect people during transmission. It was also observed that the flow ejected did not rise upwards as expected, probably due to the lack of consistent temperature and smoke generated in the system. It could also have been due to the direction of flow through the nasal cavity directed downwards.
Another important observation made by using face masks was that it could restrict the exhaled flow through the mask and keep the flow close to the body and slow down the transmission of infected aerosol particles.

Analyzing the size distributions and concentrations of aerosol particles via optical particle sizer could meet most of the requirements and stayed close to the reference ranges. The peak flow rate, velocity, travel distance, concentration distribution, and particle size were acquired from the experiments. The data processing was made easy by binning the small range particle sizes into 3 large bins categorizing them into small, medium, and large particles. The concentration of large aerosol particles were observed to reduce as they travelled farther away from the mannequin, and restricted to flow while wearing a mask. Small and medium particles were observed to have similar behavior in cases with mask and stayed close to the face region. Overall, the particle concentrations reduced with increasing distance from the mannequin.

The system requires a linear actuator working at higher speeds to control the breathing frequency and flow rate of the system much more efficiently and to reach slightly greater and more realistic peak flow rates. The hindrance of smooth flow due to narrow passages of nose and mouth could be bettered by designing nasal cavity and mouth dimension replicating human anatomy. In particular, the circular aperture in the mouth from which the air was exhaled is unrealistic and should be replaced with a more realistic opening. Apart from the design of the model, consistent thrusting of smoke into the system and improved timed breathing could yield more accurate results.

In the future, all the acquired data could be used as inputs for computational fluid dynamics analysis. For example, the framework could be applied to determine probabilities of aerosol transmission of COVID-19 in airport terminals or classrooms. However precise and accurate measures must be considered to model particle algorithm for CFD analysis.
References


