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Particle Release from Respirators, Part I: Determination of the Effect of Particle Size, Drop Height, and Load

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In late 2001, some U.S. Postal Service workers and a few members of Congress were exposed to anthrax spores. This led to an increased effort to develop employable methods to protect workers from exposure to anthrax. Some investigations focused on selection and use of respirators to protect workers against airborne anthrax. The present study evaluated the potential for several types of half-mask respirators to release deposited particles. Four brands of the most commonly used filtering facepiece respirators (hereafter termed masks) were loaded with 0.59-μm, 1.0-μm, and 1.9-μm polystyrene latex (PSL) microspheres (nominally 0.6, 1.0, and 2.0 μm) and then dropped onto a rigid surface. The load conditions were 10, 20, or 40 million particles, and drop heights were 0.15, 0.76, and 1.37 m. For the average conditions of 0.76 m, 1.15 μm size and 22 million particles loaded, the average particle release was 0.604 particles per 10,000 (95% CI: .552, .662) particles loaded for all of the filtering facepieces tested. The averaging of conditions is a useful tool to provide generalized information and is also useful when making risk estimates. For most filtering facepiece respirators, particle release tended to increase with drop height and particle size, and there appeared to be a slight inverse relationship with particle load. Two brands of reusable elastomeric half-mask respirators with P100 high-efficiency particulate air (HEPA) filter cartridges were also evaluated. Results of these tests were inconclusive. Part II in this issue addresses the release of particles when simulating removal of a filtering facepiece from a wearer's head.

Keywords filters, particles, release

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INTRODUCTION

Since the 2001 anthrax scare, there has been concern for how well respirators retain captured particles. At that time, some U.S. Postal Service (USPS) workers and a few members of the U.S. Congress were exposed to anthrax spores while at work. This resulted in an effort to determine the most effective means to protect these workers from anthrax. More recently, other threats, including avian flu, severe acute respiratory syndrome (SARS), pandemic flu, smallpox, and terrorist attacks, have come to the forefront. Protective clothing, gloves, eye and face shields, and respiratory protection have all been considered to protect the public and responders.

In the spring of 2006, the Institute of Medicine of the National Academies of Science, under contract to the National Institute for Occupational Safety and Health (NIOSH), assembled the Committee on the Development of Reusable Facemasks for Use during an Influenza Pandemic and issued a final report.[1] Since then, guidelines have been issued by various government entities on the use of respiratory protection, among other measures, by the general public and health care providers in the event of a pandemic outbreak.[2] Information gathered from the present series of studies provides important data and information on how respirators should be used and handled during such events and explores the risk that might be expected when using such devices.

Before 2001, little consideration was given to the release of particles from filtering facepiece respirators or reusable respirators used under routine handling and disposal conditions. As a result of the anthrax scare, a pilot study by Kennedy et al.[3] was conducted to investigate the conditions that influence particle release from filtering facepiece respirators. It was reported that filtering facepiece respirators loaded with 1.0-μm particles and dropped from a height of 3 ft released 0.17% to 0.29% of the loaded particles. The current study greatly expands the previous work by testing the effect of particle size, drop height, and filter load on the release of particles from filtering facepieces. An exploratory investigation of release from reusable elastomeric half-mask respirators was also conducted.

Until recently, it was believed that particles deposited in a fibrous filter would adhere to the fibers and not be dislodged easily.[4] Corn et al.[5] was the first to investigate particle adhesion. Comparison was made between particle adhesion forces and air velocities needed to re-entrain particles residing on cylindrical test surface fibers. Related research includes an investigation by Freshwater et al.[6] on the effect of particle size, gas velocity, relative humidity, and filter loading on the retention efficiency of particles larger than 5 μm on steel fibers. The authors concluded that the theory of adhesion is extremely

complex and that many factors not considered in this study also play an important role in adhesion and subsequent release, including material properties, fiber shape, and size.

Chen et al.[7] studied filter efficiency when electrostatic charge is removed from filters by looking at filters used in filtering facepieces as well as filters used in cartridge respirators. This is an important consideration since most if not all filtering facepieces made today use electrostatically charged fibers. Chen and colleagues concluded that filters treated with chemicals capable of reducing electrostatic charge caused an increase in filter penetration. Horak et al.[8] investigated the release of particles from heavily loaded HEPA filters subjected to pressure pulses equivalent to those found in tornados. Loaded filters tested with reverse airflow released large amounts of particles.

Löffler[9] studied the blow-off behavior of particles between 5 and 15 μm from fibers. It was noted that as the speed at which filtration took place was raised, higher blow-off air speeds were required to remove the trapped particles. Basic research on the filtering of different microorganisms from air has also been performed. Qian et al.[10,11] published studies of reaerosolization due to high reverse airflow (sneeze simulation) of both particles and tuberculosis bacteria from N95 disposable respirators and also from fiber filters loaded with mono- or polydisperse solid particles and liquid droplets. Qian and colleagues found that air velocity, particle size, and humidity all affect particle release from respirators. They also found that reaerosolized particles are re-entrained from the front layer (surface most proximal to the upstream airflow) of the filter.

This study provides new information previously unavailable on the fractional release of particles from contaminated respirators due to various handling procedures. The authors investigated the effect of dropping filtering facepiece and reusable elastomeric half-mask respirators that were loaded with particles. The influence of particle size, number of particles loaded onto the respirator, and drop height was evaluated. Part II, also in this issue, focuses on the release of particles resulting from the stretching of filtering facepieces to simulate removal from the user, and an estimation of risk of hazardous exposure resulting from the release of particles. The information from these studies can be used to modify workplace practices for the handling and disposal of respirators that are used to protect workers against exposure to microorganisms or other particulate toxicants.

MATERIALS AND METHODS

Selection of Particle Sizes

Particles in the 1–5 μm range are the ones considered most likely to be deposited in the alveolar region. Bacteria, some viruses, and many industrial contaminants fall into this size range.[4] Particle sizes of 0.6, 1, and 2 μm were chosen for this study as these sizes simulate many typical particles, including anthrax spores, that could enter the alveolar region.

Polystyrene latex (PSL) spheres used as the test particles have a density of 1050 kg/m^3, which may differ from the multitude of contaminants. They were chosen because they are well characterized and represent sizes and a density that are close to the expected sizes and density of many microorganisms, and they are also close to unit density, providing results very similar to their aerodynamic diameter without recalculation.

Experimental

Each experimental test comprised three stages: (1) particle loading, (2) dropping the mask onto a rigid mesh surface, and (3) particle counting using a model 3310 aerodynamic particle sizer (APS; TSI Inc., Shoreview, Minn.). Four brands of the most commonly used N95 filtering facepiece respirators were tested; they are referred to as Masks I, II, III, and IV. Given all the considered variables, there were 108 possible combinations.

To reduce the number of test runs needed and still adequately explore the effect of the defined variables, a limited test matrix was used. Tests for all variables were randomized. For each brand of disposable filtering facepiece evaluated, 1-μm particles were tested at all loading quantities (10, 20, and 40 million particles) and drop heights (0.15, 0.76, and 1.37 m). Particles sized 0.6 μm and 2 μm were tested at all drop heights but only at a loading condition of 20 million particles. A 40 million particle load was the maximum load that could be performed within a reasonable time frame. Putting load used for this experiment in perspective, 5.5×10^{11} anthrax spores is the average estimate of the number of spores per gram of highly purified anthrax.[12]

For each set of test variables, a minimum of 10 respirators were tested. Background release, hereafter termed blanks, to determine the average amount of extraneous dust that would be released and counted by the APS from unloaded filters, was also run for each test condition. More than 600 data points were obtained for this study.

In addition, two half-mask elastomeric respirators with P100 (HEPA) filters were used for this study. The respirator cartridges were loaded with 40 million 2-μm PSL spheres and dropped from a height of 1.37 m. Each brand of respirator was tested 10 times. Ten blanks were also run for each respirator. A blank consisted of a mask with unloaded (blank) cartridges and dropped from 1.37 m.

The APS uses time-of-flight measurement to count and size particles. For each of the particle sizes studied, the APS tended to place the particles into several band widths, which made it impossible to distinguish between singlets and doublets. Thus, all PSLs counted were treated as singlets, and therefore, it was important to optimize the number of singlets generated. PSL microspheres were purchased from Duke Scientific Corporation (Palo Alto, Calif.) and were suspended in water. The manufacturer ensured that all PSL spheres supplied had size distributions with coefficients of variation less than 5%; the test particles were considered to be monodisperse. The PSL sphere stock concentrations (0.6,

TABLE I. Dilutions Used to Optimize Singlet Generation

PSL Diameter	% Singlets	Coefficient of Variation (SD/% Singlets)	Dilution
0.6 μm	93.7	0.89%	1:500
1.0 μm	92.3	1.53%	1:100
2.0 μm	93.6	1.74%	1:40

1.0, 2.0 μm) are approximately 8.3×10^{10} p/cm^3, 1.8×10^{10} p/cm^3, and 2.6×10^6 p/cm^3, respectively. The concentration of spheres in the liquid suspension was adjusted to give an aerosol with a high concentration of singlets.

Various dilutions were used for generating each particle size, collected on filters and then evaluated using an optical microscope to determine the number of singlets, doublets, and triplets (Table I). Dilutions were optimized such that the percent singlets generated and their coefficients of variation were as follows: The suspension was nebulized at 35 kPa ($\sim$ 6 psig) with a flow of approximately 5.2 L/min through a DeVilbiss 45 nebulizer (Sunrise Medical, Carlsbad, Calif.) to give a loading concentration range from $\sim$31 particles/cm^3 for 2-μm particles up to $\sim$240 particles/cm^3 for 0.6-μm and 1-μm particles in the loading chamber.

The PSL particles were aerosolized in droplets of the aqueous solution, requiring complete evaporation of the water before delivery to the loading chamber. A conservative estimate of drying time, assuming a 10-μm particle, was used since it was determined that few if any droplets generated by this nebulizer were ever that large. In addition, for the drying time, a 75% relative humidity and 25.0°C were calculated, using the evaporation equation cited in Hinds,[3] to reveal that complete evaporation would require $\sim$0.11 sec, which is much less than the residence time in the neutralizer of $\sim$6.4 sec.

The surfactant content of the PSL sphere solution was checked and determined not to create any interference in the size ranges of interest. All PSL spheres used in this study were neutralized to Boltzman's equilibrium charge distribution before collection. The neutralizer used was a model 3054 (TSI Inc.) originally manufactured in November 1982 with 10 millicuries of Kr-85 (half-life = 10.3 years). The activity of the neutralizer was calculated to ensure that the particles would have adequate residence time to reach Boltzman's equilibrium. The flow rate (27 L/min) through the neutralizer was 10% lower than the calculated maximum allowed flow rate.

A breathing machine, similar to the one described by Nelson et al.,[13] was used to simulate breathing and provide the airflow through the respirators. Only the inhalation phase of the breathing cycle was used to prevent any loss of captured particles by blow-off during exhalation. This also limited turbulence within the small loading chamber. The breathing machine was operated at a light work rate (35W) with a tidal volume of 0.741 L per breath and 26 breaths per minute.

The APS provided adequate counting capabilities during both loading and release stages of each experiment. The sheath flow, nozzle flow, and total flow were checked and verified to be in accordance with manufacturer specifications. The APS was run simultaneously with a filter sample, and aerosol concentrations determined by optical microscopy

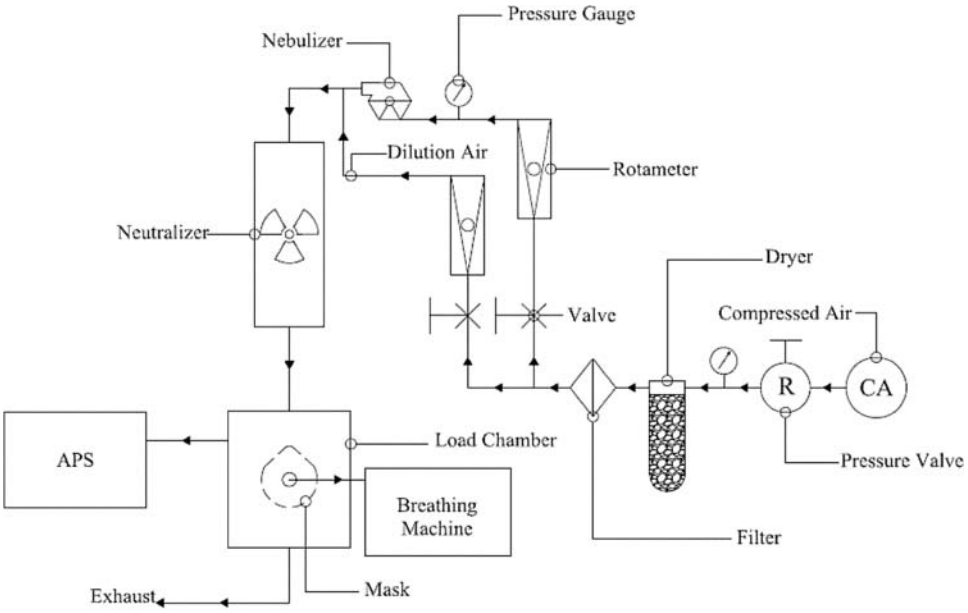

FIGURE 1. Diagram of the loading chamber and APS

were compared to the electronic APS readings and found to compare within a factor of two. Finally, each particle size was run at varying concentrations through the loading system and then read using Aerosol Instrument Management (AIM) software (TSI Inc.). It was expected that sharp peaks would correlate closely to the particle size being generated. This was confirmed. Particle loss adjustment was not made during the loading of particles because calculated losses were less than 0.04% for all particle sizes.

Loading of respirators with the test aerosol particles was done in a loading chamber (Figure 1) similar to the one described by Hinds et al.[14] In the loading chamber, each respirator was sealed to one of the contoured test fixtures specifically designed for the test respirators. The fixture was connected to the breathing machine. Particles were counted during the loading phase using the APS. They were counted for approximately 20 to 50% of the total time required to reach the target load. The APS count was used to measure the average concentration in the loading chamber, which allowed calculation of the required sample volume to reach the target loading condition.

It was assumed that the concentration did not vary significantly during the loading period. Variations in load were accounted for by using the actual load number instead of the target load level when determining fractional particle release. Once a respirator was loaded to the target concentration, it was carefully removed from the chamber and from the fixture and subjected to the treatment condition of dropping. For each run, two samples were collected: (1) an APS reading of particle concentration from the loading chamber, and (2) an APS reading of the number of released particles from the drop test. Unloaded respirators were used as blanks to correct for background particles on the filter media.

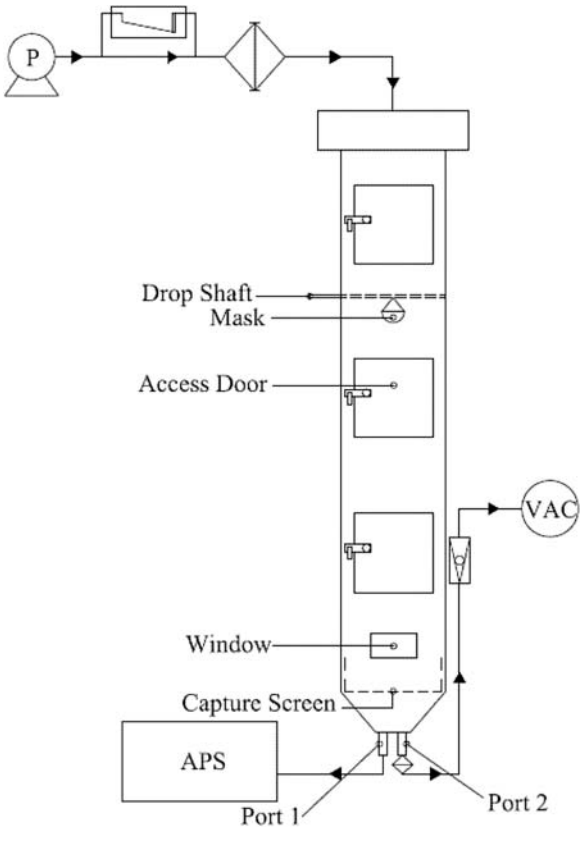

FIGURE 2. Drop chamber

The drop chamber (Figure 2) is a modified version of the chamber described by Kennedy et al.[2] The drop chamber is a vertical duct that is approximately 1.8 m long and 0.3 m in diameter. A rigid coarse screen (inverted 8-mm standard sieve) serves as the landing surface for dropped respirators. The bottom of the chamber below the screen converges and has two exhaust ports at the terminus. Port 1 was connected to the APS using flexible tubing. The flow through sampling Port 1 was governed by the APS and fixed at 5.2 L/min, providing a face velocity of 0.306 m/sec. Port 2 was added to increase flow through the drop chamber so that samples could be run more expeditiously. The flow through Port 2, which had a diameter of 50.8 mm, was adjusted so that the face velocity was equivalent to that of Port 1, providing isokinetic flow through both ports to eliminate sampling bias.

Confirmation that the velocities were isokinetic was performed by running two Portacount particle counters (TSI Inc.) simultaneously to determine that the particle concentrations registered in each port were equivalent. Correlation for isokinetic flow of the two sampling ports was high ($r = 0.99$). Total flow leaving the chamber was 42.2 L/min, which allowed the drop chamber to be cleared in less than 11 min.

To minimize contamination from room air, the top of the drop chamber was flooded with an excess of HEPA-filtered air. Air was continuously drawn down the duct during the tests so that any resuspended particles released from the respirators were removed. Excess flow was maintained at approximately 110% of the total exhaust flow. Flow rates were confirmed using a Series 2440 Kurz Thermal Anemometer (Kurz Instruments Inc., Monterey, Calif.). The chamber volume was nearly 140 L and clearance of 95% of the ambient particles required 419 L to be removed.

The chamber, once cleared of ambient contaminants, was comparable to a Class 100 cleanroom (FED-STD-209E[15]) with a maximum of $\sim$200 particles/ft^3. Clearance of the chamber was empirically verified to a background level using the APS.

APS counts of particles were adjusted for the particles removed through Port 2. Particle release per 10,000 particles loaded was then calculated.

Hose losses for each particle size were also calculated so that particle counts could be adjusted accordingly. The losses considered were diffusion loss, bend loss, and settling loss. The hose (I.D. 19 mm.) from the drop chamber to the APS was 1 m in length and had two approximately 180° bends. Calculated hose losses for particle sizes of 0.6 μm, 1 μm, and 2 μm were 0.35%, 0.87%, and 3.06% respectively. Transport

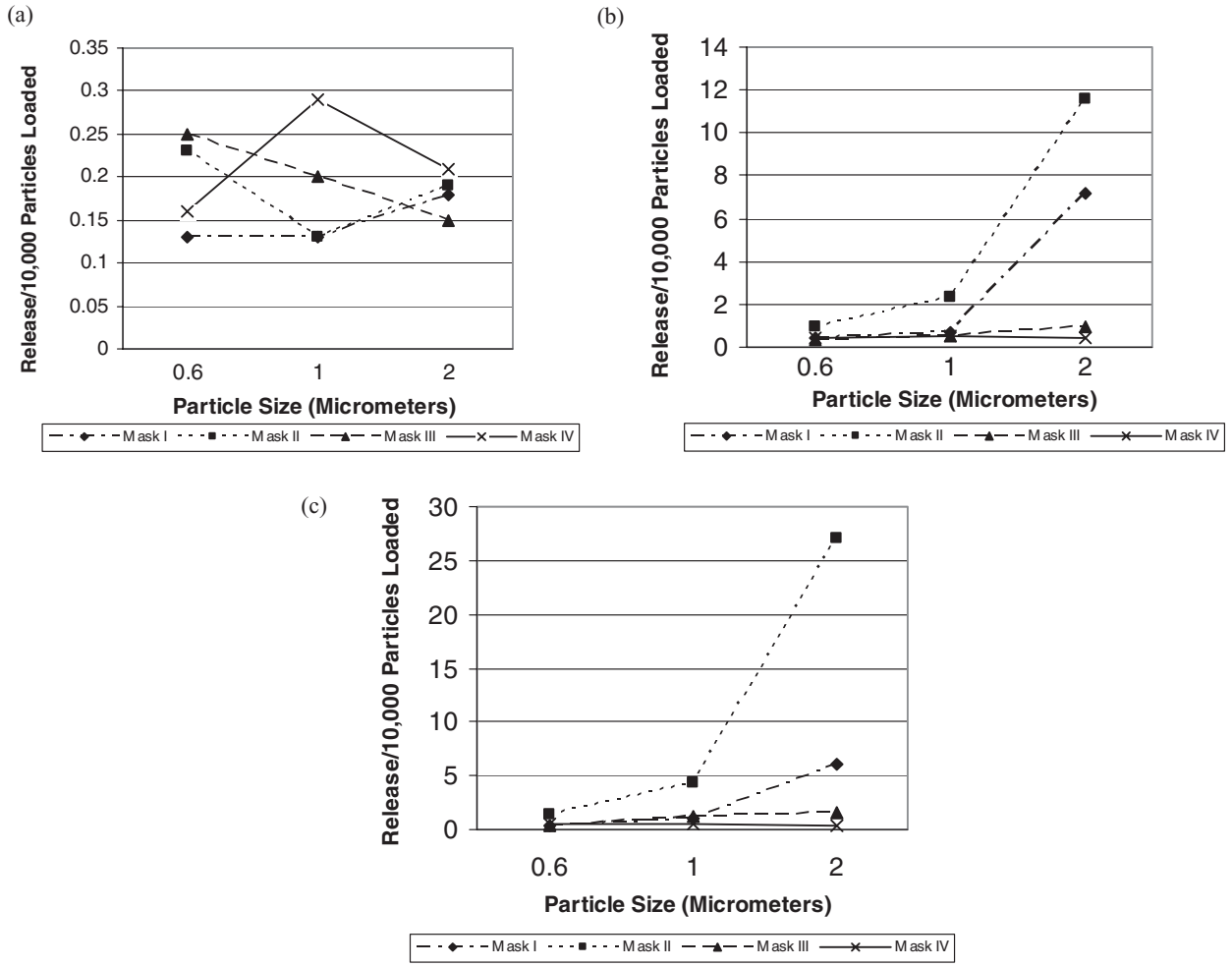

FIGURE 3. FIGURE 3 (a) Effect of particle size on fractional release at a drop height of 0.15 m; (b) Effect of particle size on fractional release at a drop height of 0.76 m; (c) Effect of particle size on fractional release at a drop height of 1.37 m

and diffusion loss equations cited in Hinds[4] were used to calculate particle losses.

All of the experiments were conducted under typical laboratory conditions; the temperature ranged from 23–25°C and the relative humidity ranged from 30–75%. Samples for filtering facepieces were discarded if the respirator did not land face down after the drop. This was done to allow maximum and consistent release for each sample. This could not be done for the reusable half-mask elastomeric facepieces.

RESULTS

Effect of Particle Size

Each brand of respirator was tested using the method described, with particle release per 10,000 loaded particles recorded for all three particle sizes. Figures 3a–c show the effect of particle size at the three different drop heights. At 0.15 m, the results show few clear trends. The one-way ANOVA results did not indicate any trends, though

TABLE II. Effect of Particle Size and Drop Height on Number of Particles Released (per 10,000) for Masks Loaded with 20 Million Particles (Log-Transformed Release Data)

Particle Size (μm)	Drop Height (m)	Mask I	Mask II	Mask III	Mask IV	P-value[A]
0.6	0.15	0.13	0.23	0.25	0.16	0.125
	0.76	0.43	0.94	0.38	0.45	0.002
	1.37	0.32	1.38	0.34	0.55	<0.001
1.0	0.15	0.13	0.13	0.20	0.29	0.001
	0.76	0.72	2.37	0.49	0.50	<0.001
	1.37	1.04	4.38	1.17	0.48	<0.001
2.0	0.15	0.18	0.19	0.15	0.21	0.896
	0.76	7.21	11.54	0.96	0.44	<0.001
	1.37	6.15	26.97	1.62	0.43	<0.001

[A]Differences of particle release between masks using ANOVA.

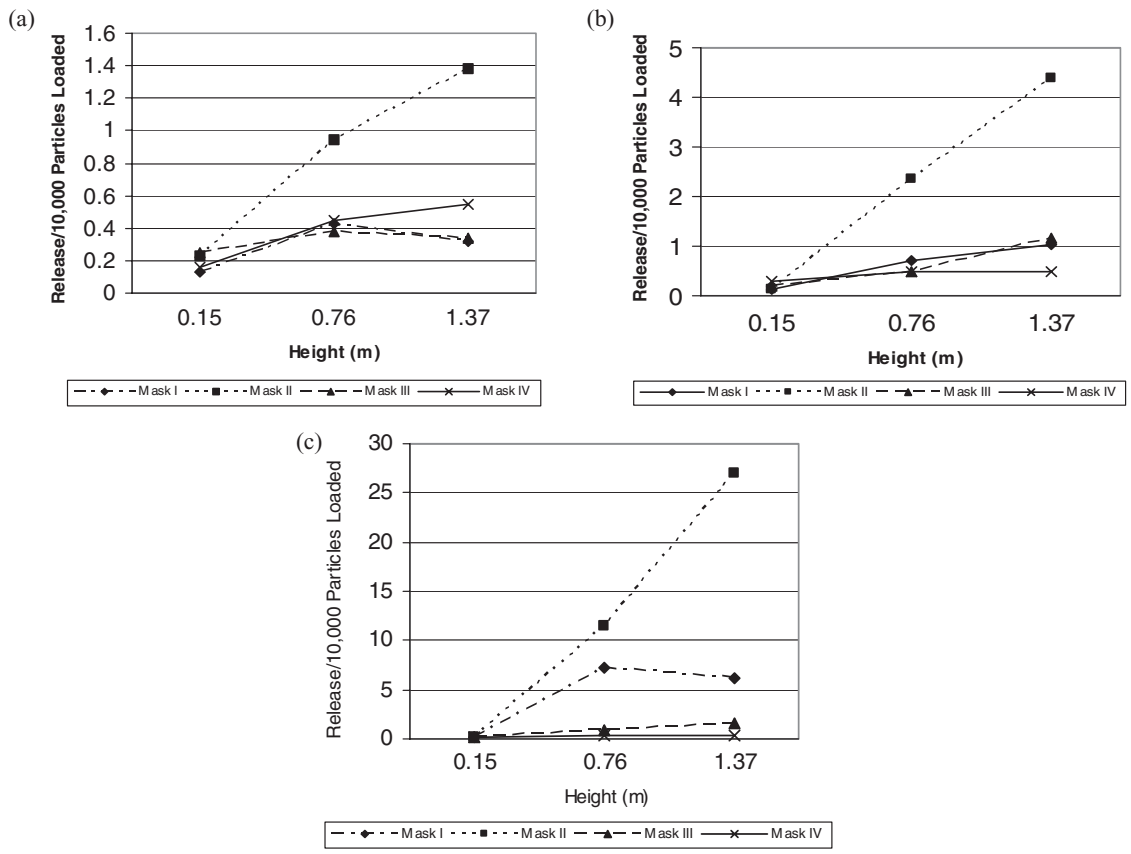

FIGURE 4. (a) Effect of drop height on fractional release of 0.6-μm particles; (b) Effect of drop height on fractional release of 1.0-μm particles; (c) Effect of drop height on fractional release of 2.0-μm particles

differences between masks can be noted. These differences were likely due to negligible release for the given height. One-way ANOVA analysis of data collected at other heights shows significant trends in terms of release by particle size. Results are summarized in Table II. The relationship readily apparent is that particle release for Masks I and II increases as particle size increases for the drop heights tested, whereas there seems to a slight increase for Mask III and a leveling off effect for Mask IV for 1.0- and 2.0-μm particles.

Effect of Drop Height

Figures 4a–c show effect of drop height for each of the three tested particle sizes. As height increases so does particle release for some masks, whereas release from others is less pronounced and appears to level off depending on particle size and height. Mask II appears to have the greatest release of particles at the higher drop distances, followed by Masks I and III. This may have been due to differences in construction or filter media. Fractional release (which describes the number of particles released by the mask as compared to the number of particles loaded onto the mask) increases with drop height between 0.15 and 0.76 m but less rapidly between 0.76 and 1.37 m and even seems to diminish for some masks. Release for Mask II continues to increase and releases significantly more particles than the other masks at 0.76 and 1.37 m.

Effect of Particle Loading

Figures 5a–c and Table III show the effect of particle load. The graphs show that a difference exists among the respirator types, depending on the drop height. At a drop height of 0.15 m there are notable differences between the respirators for the smallest load of 10 million particles. At greater drop distances of 0.76 and 1.37 m, it is clear that Mask II releases significantly more particles than the other three products. In addition, Masks I, III, and IV appear to show some trend of decreasing release as load increases.

Figure 6 shows the difference in particle release when the masks were loaded with 20 million 1.0-μm particles (at all drop heights) compared to the blank, or unloaded, condition. It is important to note that unloaded (blank) condition refers to masks that were not loaded with PSLs yet released extraneous particles, including dust that may have been attached to the masks' filter media.

Table IV is a summary of the data for the loaded and unloaded conditions for weighted means for height, particle size, and load. This is useful information as it takes all the variables into account. A denominator of 10,000 was used for blanks so that these released particles could be subtracted from the loaded masks. By using this artificial denominator, comparisons could be made to loaded masks showing the magnitude of debris naturally released from masks

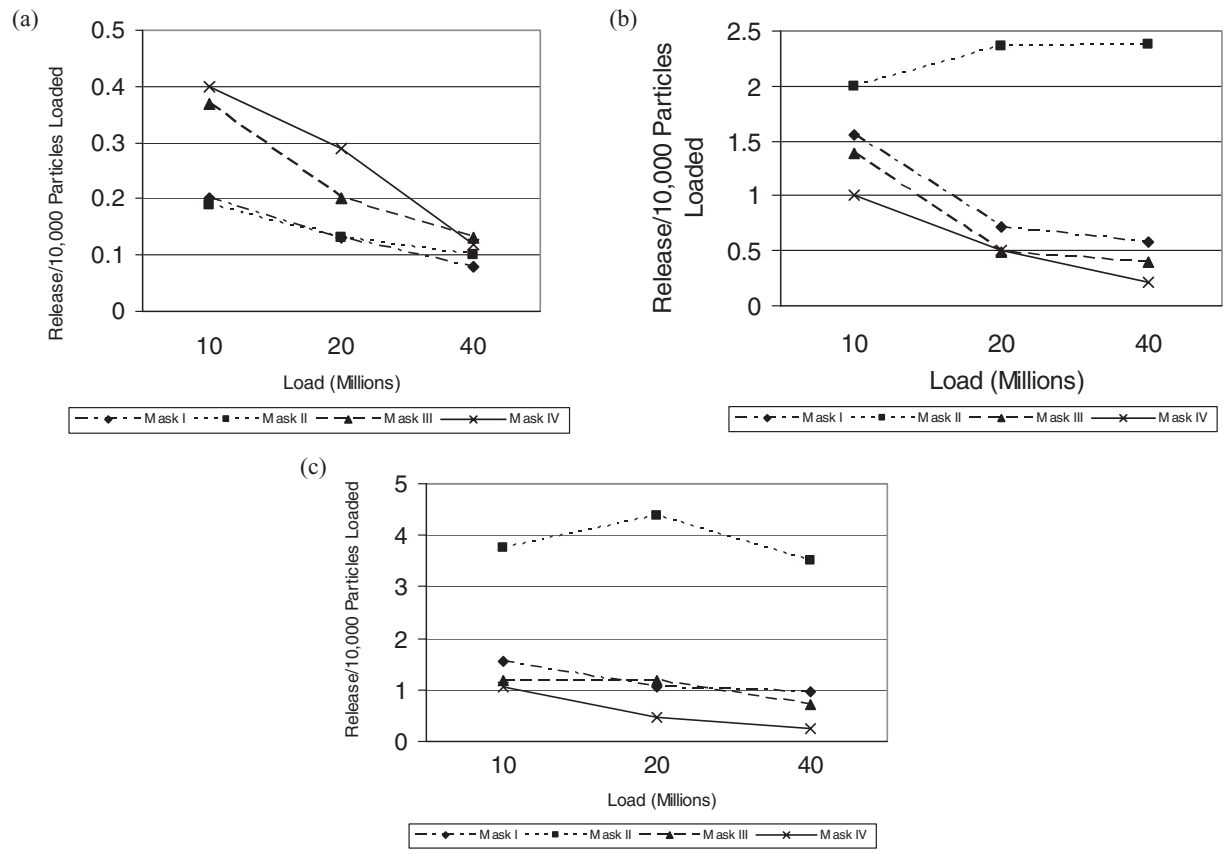

FIGURE 5. (a) Effect of load on fractional release of 1-μm particles from a drop height of 0.15 m; (b) Effect of load on fractional release of 1-μm particles from a drop height of 0.76 m; (c) Effect of load on fractional release of 1-μm particles from a drop height of 1.37 m

as compared to the release of intentionally loaded particles. It is clear that all four mask brands release more particles in the 1-μm size range when loaded than when unloaded ($p < .001$).

TABLE III. Number of Particles Released (per 10,000 loaded) for 1-μm Particles at Three Load Conditions (Log-Transformed Release Data)

Drop Height (m)	Load (millions)	Mask I	Mask II	Mask III	Mask IV	P-value[A]
0.15	10	0.20	0.19	0.37	0.40	0.016
	20	0.13	0.13	0.20	0.29	0.001
	40	0.08	0.10	0.13	0.12	0.387
0.76	10	1.55	1.99	1.38	1.00	0.123
	20	0.72	2.37	0.49	0.50	<0.001
	40	0.58	2.38	0.40	0.22	<0.001
1.37	10	1.56	3.76	1.17	1.06	<0.001
	20	1.05	4.38	1.17	0.48	<0.001
	40	0.95	3.51	0.72	0.26	<0.001

[A]Differences of particle release between masks using ANOVA.

Reusable Elastomeric Half-Mask Respirators

The results of the release of particles from reusable elastomeric half-mask respirators were considered inconclusive, and further investigation is warranted. It was noted that dropping objects of similar weight and volume in the drop chamber caused significant particle release or intrusion into the chamber. Weight and volume may be factors as well, and results must be viewed cautiously. Further investigation would appear to require experiment redesign by performing tests in a cleanroom environment to eliminate apparent particle intrusion into the chamber.

DISCUSSION

It is clear that disposable masks loaded with particles can release some of these particles under certain conditions. The mean blank-adjusted release for weighted average conditions of .76 m drop height, 1.15 μm particle diameter, and 22 million particle load between loaded masks ranged from 0.002% to 0.012%, which means that for every 20 million particles loaded on a mask, 400 to 2200 of the captured particles are released. Table II shows Mask IV released the smallest number of particles, and Mask II released the most particles followed by Masks I and III, respectively. For example, at maximum

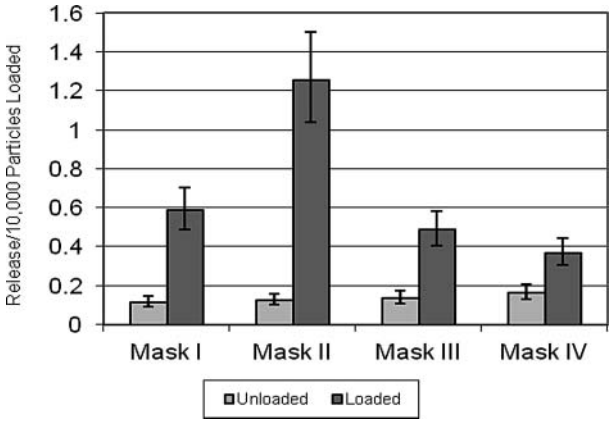

FIGURE 6. Particle release for loaded and unloaded (blank) mask (using 1-μm particles and a 20-million particle load) with 95% CIs

drop height the release rate ranged from .06% to .27%. The significant differences in release associated with the different brands of masks may be due to construction differences in the respirators or other unidentified factors, which require further investigation. The differences in release rates between this study and that of Kennedy et al.[3] may be due to various factors, including the use of electronic counting methods, the greater number of variables, and the larger sample sizes used in this study.

Another finding is that as the drop height and particle size increase, particle release often appears to increase with some leveling off in release rate for some masks. This may be due to the masks approaching their terminal settling velocity. Kennedy et al.[3] found that the filtering facepieces they tested reached a maximum velocity of 2.8 m/sec from a 0.76 m drop. It would be expected that all of the masks investigated in the present study would reach a similar velocity, as they were all of similar mass and geometry to those used previously. One would expect that at some height beyond 1.37 m a terminal velocity would be achieved, and further increases in drop height would not result in increased force at impact.

TABLE IV. Adjusted Mean and Confidence Intervals for Particle Release by Mask and Load Condition

	Mean Release (95% CI)	
Mask Type	**Unloaded**	**Loaded**
I	0.116 [0.0905, 0.149]	0.587 [0.489, 0.704]
II	0.127 [0.100, 0.160]	1.254 [1.04, 1.503]
III	0.137 [0.109, 0.172]	0.488 [0.407, 0.585]
IV	0.165 [0.131, 0.209]	0.368 [0.307, 0.442]

Notes: Adjusted mean (for average height of 0.76 m, average load 22 million, average size of 1.15 μm) and 95% CI for particle release by mask and load condition, expressed per 10,000 particle load.

An unexpected outcome is the effect of load. It appears there is a negative load effect such that as load increases particle release decreases. The original expectation was that there is enough available filter surface area that each collected particle would be able to be collected on a discrete part of a filter fiber without interference from other particles. Based on calculations done by Kennedy et al.[3] each filter has a total surface area available for collection of approximately 25,000 cm^2. Considering maximum and minimum loading conditions, this would allow between 62,500 μm^2 and 250,000 μm^2 of available fiber for each particle. The average direction of particle movement through the mask is front to back; therefore, ~50% of the filter's surface area is not available. If the load effect is found to be real, differential load due to deviant airflows should be considered, as there may be effects such as clumping of particles or some other unexplained effect.

CONCLUSIONS

The study has shown that particles are released from commonly used filtering facepiece respirators when the masks are dropped onto a hard surface. There are notable differences in release among brands, and these differences may be related to construction variations. Appropriate handling procedures should be recommended by manufacturers and must be employed by users.

The amount of particle release was affected by particle size and by drop height; in general, as either of these increased, so did particle release, although not in all cases. For 1-μm particles, there was an inverse effect of load for some of the respirators.

This study confirms that rough handling or abuse of respirators may cause particles to be released, which is contrary to the long-held belief that particles, once deposited on filters, are not easily released. Filtering facepiece respirators are used to protect workers from a wide variety of airborne hazards, including chemicals and microorganisms. Each type of contaminant has different characteristics and must be considered individually when examining the possibility of secondary contamination due to particles released from used masks.

This information should be used by employers in training their personnel on the proper care and handling of used respirators. In particular, this information should be used by personnel who handle large quantities of waste-containing discarded respirators that may harbor enough contaminant en masse to present a hazard to the workers if not appropriately dealt with. In addition, this information may be useful to regulators in formulating regulations on the use and care of respirators, as well as by respirator manufacturers in providing instructions on the proper care and handling of used respirators.

Further investigation and experimental modifications must be made to reach any conclusions as to particle release from the elastomeric half-mask facepiece respirators.

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