Size Selective Characterization and Particle Emission Rates during a Simulated Medical Laser Procedure

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THESIS
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RL
# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>CHAPTER</th>
<th>INTRODUCTION</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>..................................................................................................................</td>
<td>1</td>
</tr>
<tr>
<td>1.1</td>
<td>Background ...................................................................................................</td>
<td>1</td>
</tr>
<tr>
<td>1.1.1</td>
<td>Laser Fundamentals ...................................................................................</td>
<td>1</td>
</tr>
<tr>
<td>1.1.2</td>
<td>Pulsed Lasers Versus Continuous-Wave Lasers ..........................................</td>
<td>2</td>
</tr>
<tr>
<td>1.1.3</td>
<td>Lasing Medium .............................................................................................</td>
<td>3</td>
</tr>
<tr>
<td>1.1.4</td>
<td>Laser Wavelength .......................................................................................</td>
<td>3</td>
</tr>
<tr>
<td>1.1.5</td>
<td>Medical Laser Applications ........................................................................</td>
<td>4</td>
</tr>
<tr>
<td>1.2</td>
<td>Laser Safety Regulations .............................................................................</td>
<td>5</td>
</tr>
<tr>
<td>1.2.1</td>
<td>Laser Safety Classification Schemes ......................................................</td>
<td>5</td>
</tr>
<tr>
<td>1.2.2</td>
<td>Laser Safety Standard Requirements .......................................................</td>
<td>8</td>
</tr>
<tr>
<td>1.3</td>
<td>Summary of Beam and Non-Beam Hazards ...................................................</td>
<td>8</td>
</tr>
<tr>
<td>1.4</td>
<td>Laser-generated Air Contaminants ..............................................................</td>
<td>9</td>
</tr>
<tr>
<td>1.4.1</td>
<td>Production of Laser-Generated Air Contaminants ......................................</td>
<td>9</td>
</tr>
<tr>
<td>1.4.2</td>
<td>Sampling of Laser-Generated Air Contaminant Particles ............................</td>
<td>10</td>
</tr>
<tr>
<td>1.4.3</td>
<td>Laser-Generated Air Contaminant Particle Diameter .................................</td>
<td>12</td>
</tr>
<tr>
<td>1.4.4</td>
<td>Laser-Generated Air Contaminant Particle Shape .......................................</td>
<td>14</td>
</tr>
<tr>
<td>1.4.5</td>
<td>Laser-Generated Air Contaminant Particle Viability ....................................</td>
<td>14</td>
</tr>
<tr>
<td>1.5</td>
<td>Health Effects Associated with Exposure to Laser-Generated Air Contaminant</td>
<td>21</td>
</tr>
<tr>
<td>1.5.1</td>
<td>Health Effects from Surveys of Clinical Personnel Working with Lasers .......</td>
<td>21</td>
</tr>
<tr>
<td>1.5.2</td>
<td>Health Effects in Animal Studies ..............................................................</td>
<td>22</td>
</tr>
<tr>
<td>1.6</td>
<td>Mitigation of Laser-Generated Air Contaminant in Clinical Settings ..........</td>
<td>24</td>
</tr>
<tr>
<td>1.6.1</td>
<td>Importance of Control Strategies for Laser-Generated Air Contaminants .......</td>
<td>24</td>
</tr>
<tr>
<td>1.6.2</td>
<td>Protection of Clinical Staff from Laser-Generated Air Contaminants ..........</td>
<td>24</td>
</tr>
<tr>
<td>1.7</td>
<td>Sampling Methodologies for Laser-generated Air Contaminant Particles ......</td>
<td>27</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>CHAPTER</th>
<th>CHARACTERIZATION OF SIZE-SPECIFIC PARTICLE EMISSION RATES FOR A SIMULATED LASER SURGICAL PROCEDURE</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>2</td>
<td>Introduction ....................................................................................................................................</td>
<td>29</td>
</tr>
<tr>
<td>2.1</td>
<td>Methods ...........................................................................................................................................</td>
<td>29</td>
</tr>
<tr>
<td>2.2</td>
<td>Emission Chamber ...........................................................................................................................</td>
<td>33</td>
</tr>
<tr>
<td>2.2.1</td>
<td>Experimental Design ......................................................................................................................</td>
<td>33</td>
</tr>
<tr>
<td>2.2.2</td>
<td>Simulated Procedure ......................................................................................................................</td>
<td>37</td>
</tr>
<tr>
<td>2.3</td>
<td>Result .............................................................................................................................................</td>
<td>39</td>
</tr>
<tr>
<td>2.3.1</td>
<td>Background Measurements ...............................................................................................................</td>
<td>40</td>
</tr>
<tr>
<td>2.3.2</td>
<td>Emission Rate Results ....................................................................................................................</td>
<td>41</td>
</tr>
<tr>
<td>2.4</td>
<td>Discussion ......................................................................................................................................</td>
<td>45</td>
</tr>
</tbody>
</table>
### TABLE OF CONTENTS (continued)

<table>
<thead>
<tr>
<th>CHAPTER</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.5</td>
<td>Conclusions ................................................................. 49</td>
</tr>
<tr>
<td>2.6</td>
<td>Acknowledgements .......................................................... 50</td>
</tr>
<tr>
<td>3.</td>
<td>MICROSCOPY OF MEDICAL LASER-GENERATED PARTICLES FROM A SIMULATED SURGICAL PROCEDURE ................................................. 51</td>
</tr>
<tr>
<td>3.1</td>
<td>Introduction ........................................................................... 51</td>
</tr>
<tr>
<td>3.2</td>
<td>Methods .................................................................................... 54</td>
</tr>
<tr>
<td>3.3</td>
<td>Results ...................................................................................... 56</td>
</tr>
<tr>
<td>3.4</td>
<td>Discussion ................................................................................ 59</td>
</tr>
<tr>
<td>3.5</td>
<td>Conclusion ................................................................................ 61</td>
</tr>
<tr>
<td>3.6</td>
<td>Acknowledgements ................................................................... 62</td>
</tr>
<tr>
<td>4.</td>
<td>APPLICATION OF TWO-ZONE MODEL TO ESTIMATE MEDICAL LASER-GENERATED PARTICULATE MATTER EXPOSURES ................................................. 63</td>
</tr>
<tr>
<td>4.1</td>
<td>Introduction ............................................................................ 63</td>
</tr>
<tr>
<td>4.2</td>
<td>Methods ..................................................................................... 67</td>
</tr>
<tr>
<td>4.2.1</td>
<td>Particle Emission Rates .......................................................... 67</td>
</tr>
<tr>
<td>4.2.2</td>
<td>Two-Zone Model ........................................................................ 68</td>
</tr>
<tr>
<td>4.3</td>
<td>Results ...................................................................................... 70</td>
</tr>
<tr>
<td>4.3.1</td>
<td>Results for Simulated Operating Room ..................................... 70</td>
</tr>
<tr>
<td>4.3.2</td>
<td>Results for Simulated Treatment Room ..................................... 73</td>
</tr>
<tr>
<td>4.4</td>
<td>Discussion ................................................................................ 73</td>
</tr>
<tr>
<td>4.5</td>
<td>Conclusion ................................................................................ 77</td>
</tr>
<tr>
<td>4.6</td>
<td>Acknowledgments ...................................................................... 77</td>
</tr>
<tr>
<td>5.</td>
<td>CONCLUSIONS ............................................................................. 78</td>
</tr>
<tr>
<td>5.1</td>
<td>Experimental Design ............................................................... 78</td>
</tr>
<tr>
<td>5.2</td>
<td>Overall Results .......................................................................... 78</td>
</tr>
<tr>
<td>5.3</td>
<td>Problems identified ............................................................... 81</td>
</tr>
<tr>
<td>5.4</td>
<td>Future Research ......................................................................... 84</td>
</tr>
<tr>
<td>5.5</td>
<td>Conclusions ............................................................................... 86</td>
</tr>
<tr>
<td>6.</td>
<td>APPENDIX .................................................................................. 87</td>
</tr>
<tr>
<td>7.</td>
<td>WORKS CITED ............................................................................. 93</td>
</tr>
<tr>
<td>8.</td>
<td>VITA ......................................................................................... 103</td>
</tr>
<tr>
<td>TABLE</td>
<td>PAGE</td>
</tr>
<tr>
<td>-------</td>
<td>------</td>
</tr>
<tr>
<td>I. ANSI Z136.1 LASER CLASSIFICATION GROUPINGS</td>
<td>7</td>
</tr>
<tr>
<td>II. OPERATIONAL PARAMETERS USED FOR EXPERIMENTATION</td>
<td>38</td>
</tr>
<tr>
<td>III. BACKGROUND EMISSION RATE (PARTICLES/MIN) BY SIZE RANGE FOR BOTH SAMPLING DEVICES</td>
<td>41</td>
</tr>
<tr>
<td>IV. AVERAGE ADJUSTED EMISSION RATE FOR EACH OPERATIONAL PARAMETER COMBINATION BY PARTICLE SIZE RANGE</td>
<td>42</td>
</tr>
<tr>
<td>V. AVERAGE ADJUSTED EMISSION RATE BY PARAMETER LEVEL AND BY PARTICLE SIZE RANGE</td>
<td>44</td>
</tr>
<tr>
<td>VI. LASER PARAMETER SETTINGS USED DURING LASING FOR EACH OF FILTER SAMPLES</td>
<td>56</td>
</tr>
<tr>
<td>VII. MASS EMISSION RATE BY SIZE RANGE FOR TWO MEDICAL LASER OPERATIONAL PARAMETER SETTINGS</td>
<td>68</td>
</tr>
<tr>
<td>VIII. TWO-ZONE MODEL RESULTS IN MG/M³ FOR SIMULATED OR AND TREATMENT ROOMS AT 5, 10, 15 MINUTES AND STEADY STATE (SS)</td>
<td>72</td>
</tr>
<tr>
<td>IX. CORRECTED LGAC PERCENT SAMPLED BY PROBES</td>
<td>88</td>
</tr>
<tr>
<td>X. PERCENT OF SAMPLES WITH NEGATIVE GENERATED EMISSION RATES</td>
<td>88</td>
</tr>
</tbody>
</table>
# LIST OF FIGURES

<table>
<thead>
<tr>
<th>FIGURE</th>
<th>PAGE</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.</td>
<td>36</td>
</tr>
<tr>
<td>2.</td>
<td>41</td>
</tr>
<tr>
<td>3.</td>
<td>57</td>
</tr>
<tr>
<td>4.</td>
<td>58</td>
</tr>
<tr>
<td>5.</td>
<td>89</td>
</tr>
<tr>
<td>6.</td>
<td>90</td>
</tr>
<tr>
<td>7.</td>
<td>91</td>
</tr>
<tr>
<td>8.</td>
<td>92</td>
</tr>
</tbody>
</table>

1. Emission chamber technical drawing
2. Particles emission rate between 0.3 and 0.5 µm Dₙ for background samples by day and time
3. Background filter sample
4. SEM photographs of both laser devices and settings
5. SEM photographs of CO₂ laser filter sample using high laser settings (12 W, 5 HZ, 2.5mm)
6. SEM photographs of CO₂ laser filter sample using low laser settings (12 W, 1.2 HZ, 2.5 mm)
7. SEM photographs of Ho:YAG laser filter sample using high laser settings (12 W, 12 HZ, 1mm)
8. SEM photographs of Ho:YAG laser filter sample using low laser settings (12 W, 5 HZ, 1mm)
**LIST OF ABBREVIATIONS**

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>ACH</td>
<td>Air changes per hour</td>
</tr>
<tr>
<td>AEL</td>
<td>Accessible exposure limit</td>
</tr>
<tr>
<td>ANOVA</td>
<td>Analysis of variance</td>
</tr>
<tr>
<td>ANSI</td>
<td>American National Standards Institute</td>
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<tr>
<td>CO₂</td>
<td>Carbon dioxide</td>
</tr>
<tr>
<td>CDRH</td>
<td>Center for Devices and Radiologic Health</td>
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<tr>
<td>CW</td>
<td>Continuous wave</td>
</tr>
<tr>
<td>Dₐ</td>
<td>Aerodynamic diameter</td>
</tr>
<tr>
<td>Dₗ</td>
<td>Beam diameter</td>
</tr>
<tr>
<td>HCLS</td>
<td>Healthcare laser system</td>
</tr>
<tr>
<td>HEPA</td>
<td>High efficiency particulate air</td>
</tr>
<tr>
<td>Ho:YAG</td>
<td>Holmium yttrium aluminum garnet</td>
</tr>
<tr>
<td>HPV</td>
<td>Human papillomavirus</td>
</tr>
<tr>
<td>LEV</td>
<td>Local exhaust ventilation</td>
</tr>
<tr>
<td>LGAC</td>
<td>Laser-generated air contaminant</td>
</tr>
<tr>
<td>mJ</td>
<td>Millijoule</td>
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<tr>
<td>mW</td>
<td>Milliwatt</td>
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<tr>
<td>Nd:YAG</td>
<td>Neodymium-doped yttrium aluminum garnet</td>
</tr>
<tr>
<td>NIOSH</td>
<td>National Institute for Occupational Safety and Health</td>
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<tr>
<td>OR</td>
<td>Operating Room</td>
</tr>
<tr>
<td>OSHA</td>
<td>Occupational Safety and Health Administration</td>
</tr>
<tr>
<td>PCR</td>
<td>Polymerase chain reaction</td>
</tr>
<tr>
<td>PEL</td>
<td>Permissible exposure limit</td>
</tr>
<tr>
<td>PM</td>
<td>Particulate matter</td>
</tr>
</tbody>
</table>
LIST OF ABBREVIATIONS (continued)

PMMA  Polymethylmethacrylate
PRF   Pulse-repetition frequency
SEM   Scanning electron microscope
SUMMARY

A laboratory-based simulated surgical procedure was designed to characterize the medical laser-generated air contaminant (LGAC) particles generated during surgical procedures and to estimate exposures in theoretical rooms. Laser operational parameter settings were varied between levels to investigate the influence of parameter settings on LGAC generation.

Two medical lasers, the carbon dioxide at a wavelength of 10,600 nanometers (CO₂, λ = 10,600 nm) and the holmium yttrium aluminum garnet (Ho:YAG) laser at the wavelength of 2100 nanometers (Ho:YAG, λ = 2100 nm) were used, varying three operational parameters (beam diameter, pulse-repetition frequency [PRF], and power) between two levels and the resultant plume was collected using two real-time size selective particle counters in a laboratory emission chamber. Analysis of variance (ANOVA) was used to determine the influence of operational parameter settings on size-specific particle emission rate. Particles from a limited number of experiments were also collected on polycarbonate filters and imaged using a scanning electron microscope (SEM) in backscatter mode to study the particle characteristics and if mechanism of formation could be determined. Particles on each filter were counted and a determination on shape (irregular versus homogenous) and diameter was made. Size-specific particle emission rates were then used to demonstrate potential concentration range using a two-zone exposure model.
Results indicate power and beam diameter were statistically significant influential parameters for both lasers and for all particle size ranges, but pulse repetition frequency was only a statistically significant influential parameter for the smallest particles generated. An increase in power and decrease in beam diameter led to an increase in particle emission for the Ho:YAG laser. For the CO2 laser, higher power led to a decrease in emission rates of small particles and an increase for large particles while a smaller beam diameter led to an increase of particle emissions for most size ranges (<10μm). Beam diameter was the most influential variable in the generation of laser-generated particles at all sizes, and the three operational parameters we tested had the most influence on the generation of the smallest particle size ranges. Particle size varied, with the Ho:YAG laser producing particles in the 1–10 μm range and the CO2 laser producing particles between 1 and 50 μm in diameter. Particle shape was variable, with fibers, foam, and conglomerate particles present in our samples. Modeled concentrations for the near-field ranged between 0.03 and 0.5 mg/m³ and between 0.01 and 0.4 mg/m³ in the far-field. Results indicate concentrations in the simulated scenarios were similar to those obtained from previously reported field assessments conducted in hospital operating rooms (ORs).

The methods used in this study provide a foundation for future investigations to better estimate particle-size dependent emission rates for additional laser operational parameters in order to inform occupational exposure control strategies.
1. INTRODUCTION

1.1 Background

1.1.1 Laser Fundamentals

The American National Standards Institute (ANSI) defines the laser as a device that produces or amplifies radiant energy by stimulated emission (American National Standards Institute (ANSI) 2005). Three characteristics differentiate laser energy from other forms of energy and allow laser beam light to be focused onto areas as small as 1/100th the size of ordinary light (Rockwell Laser Industries 2007). Laser beams are monochromatic, unidirectional, and coherent. They produce light that is of a single wavelength, with waves traveling in the same direction in perfect phase either spatially, temporally or both (Rockwell Laser Industries 2007).

All lasers consist of three essential components; an active medium, an excitation mechanism, and an optical resonator. Lasers are named by either the type of medium (i.e., Argon, CO₂, neodymium-doped yttrium aluminum garnet (Nd:YAG)) or the state of the medium (i.e., gas, solid state, liquid). The medium determines the laser wavelength, and some media are able to produce a range of wavelengths that can be tuned to a specific wavelength (Rockwell Laser Industries 2007). The excitation mechanism usually consists of an energy pump, which introduces energy/light into the medium. The medium absorbs the energy and releases excited photons that in return stimulate other photons to the same excited energy level.
Optical resonators, usually mirrors, are placed on opposite ends of the medium. The rear mirror is highly reflective, while the output mirror is partially reflective, allowing some energy to escape while most is reflected back through the medium for further stimulation and amplification.

1.1.2 **Pulsed Lasers Versus Continuous-Wave Lasers**

Continuous wave (CW) lasers are devices that release energy continuously, or in pulses that are ≥ 0.25 seconds. In many tasks, continuous laser systems are inefficient in producing the desired effects, and the use of pulsed lasers may be an appropriate alternative. Pulsed lasers release short bursts of energy with a duration of <0.25 seconds, the amount of time necessary for the eyelid to close and avoid exposure to bright light (human aversion response time) (American National Standards Institute (ANSI) 2005). Pulsing increases the peak output power of the laser compared to CW lasers and can have a shockwave effect in tissue or other material. The short pulse length and increased peak power creates an added risk to humans, who may be exposed to the beam (Pierce et al. 2011, 447–466).

There are two types of pulsed lasers: Q-switch and Mode-locked. Q-switch lasers have pulse durations between 1 nano-second and 100 nano-seconds (Rockwell Laser Industries 2007). Mode-locked lasers have ultra-short pulses of less than 1 nano-second. Mode-locked lasers are used in corneal surgery or fluorescence microscopy while Q-switched lasers are utilized for tattoo removal and metal cutting (Rockwell Laser Industries 2007).
1.1.3 **Lasing Medium**

Laser types are classified by the medium used for exciting the electrons into producing specific wavelengths, and can either be a solid, gas, or dye. Solid state lasers were the first type of lasers invented and are common because they maintain a stable single frequency and have a wide range of power level capability. They can have power levels as high as 5 kilowatts (kW) for cutting or welding, or as low as 1 watt (W) for analytical chemistry devices (Koechner 2006, 38–101). Dye lasers are the least prevalent laser type since they degrade quickly with use, and the organic dyes are hazardous and difficult to handle, but they have a broad wavelength tenability (Paschotta 2008), allowing the laser to be tuned to a range of wavelengths, which may be useful in research applications. Gas lasers are the most common type available, have a long life span, and no degradation with use. They are available at almost all wavelengths and can be scaled to size with little added cost (Masamori and Walter 2007).

1.1.4 **Laser Wavelength**

The effect laser radiation has on tissue is dependent on the wavelength and on the composition of the tissue. Wavelengths between 100 and 380 nm are best absorbed by proteins in tissue and create a dominant photochemical effect that breaks intermolecular bonds, denatures proteins, and at high irradiance can incise tissue. Wavelengths between 400 and 780 nm are absorbed by melanin, hemoglobin, and calcium in bone and can produce both thermal and photochemical effects. Lasers in the near-infrared range (780–3,000 nm) produce photo-disruptive effects and can induce photocoagulation while lasers that operate in the far-infrared
Lasers used in healthcare applications are referred to as Healthcare Laser Systems (HCLS), and include the laser, a system that can direct the output energy, power supply, and other components. Four are recognized as the primary clinical/surgical HCLS and are the most widespread: Argon, Nd:YAG, Ho:YAG, and CO₂ laser systems (American National Standards Institute (ANSI) 2005).

Laser systems are useful in the clinical setting because they produce three main desirable effects in human tissue: thermal, ionizing, and photochemical. The thermal effect of lasers is used to seal, cauterize, or vaporize tissue for incising or excising. Laser systems are preferred over other methods because they can cut and seal simultaneously; in some cases they can eliminating the need for sutures and reducing bleeding and decrease recovery time (American National Standards Institute (ANSI) 2005; Fornaini et al. 2007, 381–392). Ionization effect is used to treat glaucoma and to open posterior capsules that have become opacified after cataract extraction. The effect is created by having short pulses with high peak power, which creates an acoustic shockwave effect creating microscopic holes in tissue (American National Standards Institute (ANSI) 2005). The photochemical effect is used to create cuts without notable necrosis, or to promote tissue regeneration (Huang et al. 2011, 602–618).

In clinical specialties, HCLS have many uses. In ophthalmology, they are used in the surgical management of proliferative diabetic retinopathy and in age-related

(3,000–10,000 nm) can incise, excise, and vaporize tissue (Council on Scientific Affairs 1986, 900–907).
maculopathy to repair retinal breaks and to incise vitreous structures. In
dermatology and plastic surgery, HCLS are used because they provide rapid wound
healing. They are used in the treatment of port wine stains, in the incision and
cauterization of excising vascular tumors and lesions, in removal of keloids, in tissue
vaporization, and in tattoo and wart removal. In general surgery HCLS are used for
excising and incising by vaporization, including the excision of decubitus ulcers,
removal of leukoplakia and premalignant lesions of the oral cavity, with patients
that have blood clotting problems, and for palliative excision of malignant diseases.
In gynecology they are used in the treatment of cervical intraepithelial neoplasia, in
laser excisional conization, and in the treatment of vaginal and vulvar intraepithelial
neoplasia. In neurosurgery HCLS are used to remove or incise tissues that are near
sensitive neural or vascular structures and to remove malignant tumors. In
otolaryngology they are used in the treatment and management of otosclerosis,
rhinology, and hereditary telangiectasia and in facial plastic surgeries for the

1.2 Laser Safety Regulations

1.2.1 Laser Safety Classification Schemes

Lasers in the United States are regulated by the US Federal Drug
Administration's (FDA) Center for Devices and Radiological Health (CDRH) and
require laser manufacturers to test their devices prior to reaching the market and to
provide a hazard classification for the device using criteria from the CDRH or using
the ANSI standard for safe use of lasers Z136.1. The two organizations (CDRH and
ANSI) employ different hazard classification groupings, but ANSI follows an
international standard that is more widely adopted than the CDRH system. The hazard classification is based on the accessible exposure limit (AEL), the maximum emission permitted within a laser class, and is dependent on the wavelength, laser mode (pulsed versus continuous) and power (American National Standards Institute (ANSI) 2007).

Lasers with a hazard classification of Class 1 or 1M are the least likely to cause injury and have a maximum exposure duration of no more than 30,000 seconds if wavelength is <0.7 µm, or 100 seconds if wavelength is greater than 0.7 µm. Class 2 and 2M lasers are in the visible spectrum and have an AEL greater than Class 1 and 1M lasers, but produce an accessible average radiant power of less than 1 milliwatt (mW). Class 3R lasers are classified as having an AEL between 1 and 5 times that of a Class 1 AEL for wavelengths less than 0.4 µm or greater than 0.7 µm, or less than five times the Class 2 AEL for wavelengths between 0.4 and 0.7 µm. Class 3B lasers operate outside of the retinal hazard region (<0.4 and >1.4 µm) and have an AEL greater than Class 3R lasers, but an average radiant power less than 0.5W for continuous lasers, or 0.125 joules for pulsed lasers. Class 4 lasers have an AEL greater than Class 3B lasers, and are considered the most hazardous (American National Standards Institute (ANSI) 2007). Most medical lasers do not contain any device that limits the AEL and usually acquire the highest hazard classification. The hazard classification scheme is presented in Table I.
<table>
<thead>
<tr>
<th>Laser class</th>
<th>Description</th>
<th>Power limit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Class 1</td>
<td>Incapable of causing injury during normal operation</td>
<td>0.04 mW (for blue)</td>
</tr>
<tr>
<td>Class 1M</td>
<td>Lasers emitting between 302.5 nm and 4000 nm and are safe except when used with optical aids</td>
<td>0.04 mW</td>
</tr>
<tr>
<td>Class 2</td>
<td>Lasers emitting between 400 nm and 700 nm (visible range) and incapable of causing injury in 0.25 seconds of viewing.</td>
<td>1 mW</td>
</tr>
<tr>
<td>Class 2M</td>
<td>Lasers emitting between 400 nm and 700 nm and that are hazardous when viewed through an optical instrument.</td>
<td>1 mW</td>
</tr>
<tr>
<td>Class 3R</td>
<td>Lasers that are marginally unsafe for intrabeam viewing and that have an AEL within 5 times class 1 limits for invisible lasers</td>
<td>1–5 mW (for CW lasers)</td>
</tr>
<tr>
<td>Class 3B</td>
<td>Lasers that are hazardous during intrabeam viewing, but normally safe during defused reflections</td>
<td>500 mW (for CW lasers)</td>
</tr>
<tr>
<td>Class 4</td>
<td>Lasers that are both eye and skin hazards during both defused and direct viewing and that are a potential fire risk.</td>
<td>above 500 mW</td>
</tr>
</tbody>
</table>
1.2.2 **Laser Safety Standard Requirements**

Safe use of lasers in health care facilities is detailed in the ANSI Z136.3 Safe Use of Lasers in Healthcare Facilities Standard. The standard requires the appointment of a laser safety officer (LSO) that oversees the creation and implementation of a laser safety program in the workplace. Responsibilities of the LSO include but are not limited to correctly classifying laser systems, implementing and/or overseeing standard operating procedures (SOP), scheduling maintenance and servicing of HCLS, implementing procedural controls, ensuring proper labeling and signage on laser system and on entryways, guaranteeing proper guarding of laser equipment, and protecting workers and patients by providing proper eyewear and a safe and secure laser procedure room (American National Standards Institute (ANSI) 2005).

The US Occupational Safety and Health Administration (OSHA) does not have a laser specific standard. Instead, the agency relies on the general duty clause and on ANSI standards and FDA/CDRH laser manufacturer requirements to enforce safety and health standards in the workplace.

1.3 **Summary of Beam and Non-Beam Hazards**

Laser hazards are divided into two subgroups: beam and non-beam hazards. Beam hazards include skin and eye injuries and are caused by the absorption of the electromagnetic energy waves and conversion of the energy into heat as well as other forms of energy (Rockwell et al. 2008). The absorption of energy by skin and tissues is dependent on tissue characteristics as well as laser settings and
Injuries can include protein denaturing and thermal burns. The eye is considered the most vulnerable organ to laser radiation, as it has limited protective structures that would reduce the absorption and adverse health effects. Eye injuries can include retinal and corneal burns as well as cataract formation (Rockwell et al. 2008).

Non-beam hazards include electrocution, fire, trips, and slips and can be caused by the absence of a laser safety program or lack of training or enforcement. In clinical procedures the interaction between a laser beam and the target matter generates a smoke or plume, or LGAC. The generation of LGAC is mostly associated with the use of class 3B and 4 laser systems. Every year more than 500,000 workers, including surgeons, nurses, anesthesiologists, and surgical technologists, are exposed to laser or electrosurgical smoke (Occupational Safety and Health Administration (OSHA) 2008), an exposure that has not been well characterized.

1.4 Laser-Generated Air Contaminants

1.4.1 Production of Laser-Generated Air Contaminants

Production of LGAC occurs when cell membranes are heated and undergo pyrolysis and combustion or rupture and release steam and cell contents (Alp et al. 2006, 1–5). The aerosolized plume is composed mostly of water (95%), with the remaining 5% composed of gas-phase contaminants and particulate matter that may contain viruses, bacteria, blood, and tissue particles (Gonzalez-Bayon, Gonzalez-Moreno, and Ortega-Perez 2006, 619) that may aerosolize, remain viable, and infect healthcare workers (Calero and Brusis 2003, 790–793; Hallmo and Naess 1991, 425–427). The smallest particles in the plume are produced from the uniform
evaporation of liquids and are generally spherical in shape. Dispersive spectrometry and scanning electron microscopy determined the smallest particles are mostly comprised of sodium, potassium, magnesium, calcium, and iron, while larger particles were produced mainly from the mechanical ejection of tissue fragments are comprised of carbon and oxygen (Weld et al. 2007, 347; Descoteaux et al. 1996, 152; Vanderpool and Rubow 1998). It has been suggested that particle ejection (or “splatter”) is dependent in part on the power, wavelength, and PRF of the laser (Canestri 1999, 199–203). Canestri et al. (1999) demonstrated that lower emission frequencies (wavelengths) led to greater spatial separation of smoke, while higher power densities led to higher particle ejection speeds and deposition farther from the lasing area (Canestri 1999, 199–203).

1.4.2 Sampling of Laser-Generated Air Contaminant Particles

A number of studies collected particles during laser and electrocautery surgical procedures. Kunachak et al. (1998) and Taravella et al. (2001) performed air sampling by placing a filter at the inlet hose of a smoke evacuator during laser surgical procedures (Kunachak and Sobhon 1998, 278–282; Taravella et al. 2001, 604–607) and analyzing filters with a SEM. Kunachak et al. (1998) analyzed filters by randomly choosing five areas on each filter and calculating the range and mean of the particle diameters and average particle density of all particles (Kunachak and Sobhon 1998, 278–282). Taravella et al. (2001) examined the particles on the filter by choosing two central areas on each filter with the most particles and used the US National Institute of Health (NIH) Image software to identify particles using defining contrast parameters. The software was used to best-fit particle images to a circle.
and to calculate the areas occupied by the circles; the values were then converted to
diameters in microns (Taravella et al. 2001, 604–607). Two other studies collected
particles on size-selective filters using multistage cascade impactors and measured
the change in mass of each filter pre- and post- sampling (Nezhat et al. 1987, 376–
382; Freitag et al. 1987, 283–288). Derrick et al. (2006) generated saline particles
with a mean diameter of 0.04 µm using a TSI particle generator and tested the
ability of surgical masks to filter ultrafine particles by counting particles able to
cross the filter media using a portable particle counter (Portacount) with a
collection size range of 0.02 to 1 µm (Derrick et al. 2006, 278–281). Another study
collected particles using a condensation particle counter (CPC, model 3007, TSI)
with a size range between 0.01 and 1 µm and at a maximum detection limit of
10,000 particles per cm³ of air (Brüske-Hohlfeld 2008, 31).

Collection of particles using filter media was reported as the most common
method employed to collect LGAC particles for counting and sizing (Kunachak and
Freitag et al. 1987, 283–288). The method is an efficient collection method, but
analysis of filters has limitations that may affect final results. When analyzing using
a SEM, counting or sizing irregularly shaped particles can skew the final results.
Filters must also have low concentrations of particles in order to facilitate counting.
Weighing of filters removes the need for low particle concentrations, but the
sensitivity of the method only allows for the estimation of particles at high
concentrations and at large diameters. Real-time particle counters are able to count
and size LGAC particles at large size-ranges; but the method may be limited when
the particles are wet, irregularly shaped and when they are produced at high concentration levels. The best method when attempting to determine the characteristics of the LGAC plume may be a combination of methods to validate results and to provide additional information including shape characteristics and mechanism of formation.

1.4.3 **Laser-Generated Air Contaminant Particle Diameter**

Aerodynamic diameter (Dₐ) standardizes particles into unit density spheres based on their settling velocity (Hinds C 1999) and is a determinant of deposition of particles in the lungs (Shapiro et al. 1991, 123–134; Davies, Heyder, and Subba Ramu 1972, 591–600). It is believed that the transfer of viable cellular or viral material through the LGAC plume would be a function of particle size, most likely associated with larger particles. Ultrafine particles (<1 μm in Dₐ) are known to deposit deep in the lungs and may transfer directly into systemic circulation and affect secondary target organs (Kreyling et al. 2002, 1513–1530; Takenaka et al. 2001, 547–551; Oberdorster et al. 1995, 111–124; Nemmar et al. 2001, 1665–1668; Nemmar et al. 2002, 998–1004; Nemmar, Hoet, and Nemery 2006, A211-2; author reply A212-3).

Studies have reported variable size ranges and mean diameter of LGAC particles. Alp et al. (2006) found that particles generated using electrocautery devices are smaller (0.1 μm) than those produced during medical laser procedures (~0.3 μm) (Alp et al. 2006, 1–5). Kunachak et al. (1998) reported a larger mean laser-generated particle diameter by irradiating laryngeal papillomas with a CW CO₂ laser at 10 W and collecting the laser-generated plume using 0.45 μm pore size
Micropore filters. The filters were attached to the tip of a smoke evacuator hose and the evacuator was set to the maximum flow rate during the procedure. A scanning electron microscope was used to count and estimate particle $D_a$. The authors measured particles ranging in size between 0.5 and 26 $\mu$m, with 80% being 0.8 $\mu$m $D_a$ (Kunachak and Sobhon 1998, 278–282). Freitag et al. (1987) performed experiments by irradiating 1 cm$^3$ of sheep bronchial tissue using an Nd:YAG laser between 15 and 75 W. A seven-stage cascade impactor was used to collect the sample and each stage was weighed to determine the mass concentration. The mean $D_a$ was determined to be 0.54 $\mu$m and mass per cubic meter of plume was 920 mg (Freitag et al. 1987, 283–288). Nezhat et al. (1987) collected laser plumes during 17 separate CO$_2$ laser treatments of endometriosis; the laser was used with a beam diameter of 0.5 mm and power between 15 and 30 W. Samples were collected in sterilized plastic bags that were subsequently sampled using a Marple personal cascade impactor for three minutes. The smallest stage of the impactor was designed to collect particles >0.2 $\mu$m. The authors determined a particle size distribution between 0.1 and 0.8 $\mu$m using a SEM, with a median $D_a$ of 0.31 $\mu$m (Nezhat et al. 1987, 376–382). Hahn et al. (1995) analyzed particles generated during Nd:YAG laser corneal ablation using light scattering technology and measured particle diameters at increased height from the plume. An average particle diameter of 0.15 $\mu$m was measured 150 $\mu$m from the corneal surface and 0.11 $\mu$m was measured at 720 $\mu$m from the surface (Hahn, Edinger, and Pettit 1995).
1.4.4 Laser-Generated Air Contaminant Particle Shape

Particle size and shape are the main determinants of location of deposition in the human lung. Shape influences a drag force that affects the transport properties of particles (McDonald 2004, 1069) and influences the light–scattering properties used to determine particle diameter by some particle sizers/counters (McDonald 2004, 1069).

A limited number of studies have reported medical laser-generated particle shape. Weld et al. (2007) collected particles from incisions of porcine muscle tissue using four different energy-based instruments. Particles were collected on electron microscope grids by electrical field deposition. A SEM (settings not reported) was used to observe two distinct particle shapes, small homogenous spheres and large irregular fragments (Weld et al. 2007, 347). Descoteaux et al. (1996) collected particles on aluminum substrate foils using a seven–stage cascade impactor during laparoscopic surgeries on animals and humans. Sixty photographs were randomly taken and examined using a SEM and two morphologically different particle distributions were observed: larger particles (>10 µm in diameter), described as being heterogeneous and resembling tissue fragments, suggesting that they may be cellular components; and small particles described as being homogenous spheres (Descoteaux et al. 1996, 152).

1.4.5 Laser-Generated Air Contaminant Particle Viability

Viable cellular and viral material in the laser-generated plume is a concern that is most likely associated with larger sized particles. Two case studies have suggested that the aerosolization of viable material and subsequent infection of
healthcare staff is possible (Calero and Brusis 2003, 790–793; Hallmo and Naess 1991, 425–427). The first report involved a laser surgeon who developed laryngeal papillomatosis of a type known to be associated with anogenital condylomas. He had performed laser treatments using an Nd:YAG laser on cancers of the distal colon and rectum of some patients known to have anogenital condyloma acuminata. During the procedures, the staff used eye protection, conventional masks, and gloves; Endoscopic smoke evacuation was not used (Hallmo and Naess 1991, 425–427). The second case involved a gynecological OR nurse who routinely assisted in the excisions of anogenital condylomas and who developed laryngeal papillomatosis. Further investigation determined the development of the papillomatosis was most likely caused by an occupational exposure (Calero and Brusis 2003, 790–793).

Two studies were able to collect viable cellular material from aerosolized LGAC during medical laser procedures. Capizzi et al. (1998) collected LGAC during 13 CO₂ laser resurfacing procedures on Millipore high efficiency particulate air (HEPA) filters attached to a smoke evacuator. The laser was operated at an energy fluence of 500 mJ/cm², two five-minute samples were collected during each procedure. Two bacterial and two viral cultures were sampled from each filter and incubated for a maximum of 14 and 28 days, respectively. Five of thirteen cultures grew coagulase-negative Staphylococcus; one also grew Corynebacterium and another grew Neisseria; none of the viral cultures were positive (Capizzi, Clay, and Battey 1998, 172–174). Ferenczy et al. (1990) collected air samples during CO₂ vaporization treatments of condylomata acuminatas and acetowhite condylomatas from 110 patients with diagnosed human papillomavirus (HPV) infection. Dacron
swabs were used to collect samples from pre-filter canisters, vacuum tubes, and from the nasopharynx, eyes, and ears of the surgeon. The swabs were analyzed using ViraType HPV DNA Typing Kit. Prior to treatment, swab samples identified HPV DNA in 65 of 110 condylomatas and neoplasias that were to be removed. After treatment, 20% of pre-filter canisters tested positive for HPV DNA, but no swabs from surgeons were positive (Ferenczy, Bergeron, and Richart 1990, 1271–1274).

Other studies have not detected any viable material in the laser plume. Bellina et al. (1982) sampled during irradiation of condylomata acuminatas lesions and collected the laser plume on 25 mm Millipore filters with a pore size of 0.8 µm using an inline vacuum system. The filters were re-suspended in solution for extraction of organic acids, Glucose 1-C, and RNA and DNA analysis. No viable DNA with oncogenic potential was detected (Bellina, Stjernholm, and Kurpel 1982, 268–270). Another study treated HPV positive warts of five patients using an Erbium-doped yttrium aluminum garnet laser at a beam diameter of 2mm, energy of 175 mJ and a PRF of 5 Hz. Samples were collected on a cotton swab from the laser hand piece and were analyzed for HPV DNA using polymerase chain reaction (PCR) and results were compared to samples of the wart prior to ablation. Viable HPV-2 strains in the collected material were detected, but not the strain contained in the wart of patients, indicating that the treatment did not release any viable HPV (Hughes and Hughes 1998, 426–428).

Laboratory-based efforts have also investigated the ability of particles to remain viable and to cause infections. Byrne et al. (1987) lased agar media tubes containing *Staphylococcus aureus* and *Escherichia coli* with a CO₂ laser in pulsed and
continuous mode and at variable power levels between 6 and 45 W with a beam diameter of 0.3 mm. Samples were collected using silastic tubing attached to a Casella slit sampler and analyzed by washing out the inside of the tubing into agar media. Viable bacteria were detected in all instances and \textit{Staphylococcus} was found to be more resistant to thermal effects than \textit{Escherichia coli} (Byrne et al. 1987, 265–273). Wisniewski et al. (1990) used a CO\textsubscript{2} laser with a power density of 666 W/cm\textsuperscript{2} to irradiate HPV positive genital lesions and collected the particles on the inner surface of a black metal cylinder placed around the tissue. Particles removed from the cylinder surface were between 100 and 200 µm and electron microscopy revealed nucleate keratinized squamous cells, but insufficient DNA was collected to determine if HPV was present in the sample (Wisniewski et al. 1990, 1117–1123). A third study simulated the vaporization of corneal tissue by lasing culture plates infected with a porcine herpes virus strain using an excimer laser at 150 and 180 mJ/cm\textsuperscript{2} and beam diameter of 4 mm. Samples were collected on uninfected culture plates placed in an inverted position above the treatment site and were observed daily for four days. Examination of the uninfected culture plates post sampling contained visible cellular debris, but none of the culture plates developed a viral infection (Hagen et al. 1997, 206–211). A separate study irradiated tissue pellets infected with HIV using a CO\textsubscript{2} laser at a beam diameter at 1.5 and 2.5 cm, power of 20 W, and fluence of 500 W/cm\textsuperscript{2}. Samples were collected by suctioning plume through clear silastic tubing then bubbling through a flask with sterile culture medium. The tubing and culture medium were incubated for either 14 or 28 days and tested weekly for viability. At the end of the incubation period, PCR was used to
test for the p24 protein, a determinant of the HIV virus. No HIV virus was detected
in the flask media, but cultures from the tubing were positive for HIV virus in 3 of 12
tube samples after one week, and in 1 of 12 samples after two weeks; no samples
remained viable after 28 days. PCR analysis was positive for pro-viral DNA in all
samples from the silastic tubing immediately after sampling and at 14 days after
sample collection (Baggish et al. 1991, 197–203).

Garden et al. have written two papers regarding the infectivity of the plume.
In 1988 they used a CO$_2$ laser at 4 and 12 watts, beam diameter of 0.2 and 2 mm,
and in either continuous or pulsed mode to vaporize papillomavirus-infected
verrucae. The generated plume was collected in an in-line chamber using a
phosphate-buffered saline solution. Subsequent analysis detected HPV DNA in two
of seven samples vaporized (Garden et al. 1988, 1199–1202). In 2002 they
attempted to simulate infectivity by exposing samples of bovine fibropapillomas
that were positive for bovine papillomavirus (BPV) to a CO$_2$ laser at power settings
of 12, 8, and 4 W, beam diameter of 2.0 mm, and in continuous and pulsed mode.
The generated laser plume was collected in an in-line chamber using a phosphate-
buffered saline solution and was analyzed for papillomavirus and later inoculated
into bovine calves. Analysis of the sampled solution detected papillomavirus DNA in
all samples, and calves developed tumors at sites of inoculation indicating that the
plume was viable and able to transmit virus (Garden et al. 2002, 1303–1307).

The latest research into the infectivity of the plume occurred in 1996 when
Kunachak et al. (1996) collected papilloma tissue from patients with known cases of
recurrent respiratory papillomas. Tissues were lased with a CO$_2$ laser set at 10 W,
beam diameter of 0.5 mm, and power density of 1667 W/cm². The plume was collected using a suction tube with an inline 0.22 µm pore-size filter and plated onto growth culture media. Growth in the cultures was observed for 45 days. Cell growth was not present in any of the filter samples suggesting that laryngeal papillomas are unable to survive continuous CO2 laser irradiation (Kunachak, Sithisarn, and Kulapaditharom 1996, 1031–1033).

The infectivity of the laser plume has also been simulated by the purposeful infection and subsequent lasing of tissue. In one study postmortem human skin was injected with gram positive bacteria spores and then irradiated using a CO2 laser at 5, 10, and 20 W in continuous and pulsed modes with pulse length of 0.05, 0.2, and 0.2 seconds. The generated laser plume was collected using an impinger and Casella slit sampler and cultured for 48 hours prior to counting the bacterial colonies on each plate. Microscopy of cultures plates found some spore containing particles at low irradiance levels and no spores at high irradiance (>750 W/cm²) levels (Matthews, Newsom, and Walker 1985, 230–233). McKinley & Ludlow (1994) injected Escherichia coli into the root canal of five freshly extracted single-rooted human teeth and lased the canals with an argon laser at 488 and 514 nm and power at 2 W and 5 Hz. Agar plates were inverted above the lasing site to collect the laser-generated plume and incubated for 24 hours. Subsequent analysis determined all five sample plates were positive for the same Escherichia coli strain injected into root canal samples (McKinley and Ludlow 1994, 558–559). Mullarky et al. (1985) injected Escherichia coli and Staphylococcus into blocks of shaved and disinfected porcine tissue and lased the tissue using a CO2 laser at 25 W with either a defocused
(3mm) or focused (0.2 mm) beam diameter. The generated laser plume was collected in an in-line chamber using a sterile solution that was later passed through a membrane filter and incubated for four days. Bacterial growth was observed daily and gram positive bacteria were found to be more resilient compared to gram negative bacteria and the defocused beam seemed to aerosolize less viable bacteria compared to the focused beam (Mullarky, Norris, and Goldberg 1985, 186–187).

Examination into the viability of the laser-generated plume indicates the plume can contain viable bacteria and virus and may cause infection. The ability to aerosolize viable material may be partially dependent on the wavelength of the laser, as certain wavelengths are absorbed better in tissue and increase the temperature of the plume, causing rapid heating and vaporization (Hughes and Hughes 1998, 426–428). Some have suggested low power levels create higher numbers of viable particles (Matthews, Newsom, and Walker 1985, 230–233), while other work suggests high power densities can kill viable material in the generated laser plume (Kunachak, Sithisarn, and Kulapaditharom 1996, 1031–1033). There currently is not an understanding of factors that lead to the viability of the laser-generated plume. Most studies have only investigated the CO₂ laser in quantitative studies and at a limited number of operational parameter settings. Medical laser operators may benefit from an investigation of multiple wavelengths and operational parameters to determine the levels and wavelengths that are most likely to aerosolize viable material.
1.5 **Health Effects Associated with Exposure to Laser-Generated Air Contaminant**

1.5.1 **Health Effects from Surveys of Clinical Personnel Working with Lasers**

Health effects associated from exposure to LGAC are based on limited findings from animal studies and surveys of clinicians. Health effects reported from a survey of 11 workers involved in medical laser procedures did not report any health effects related to LGAC; however some did complain about the smell and also about vision impairment from the procedures (National Institute for Occupational Safety and Health (NIOSH) 1990). Lobraico et al. (1989) surveyed 4,500 laser users in multiple clinical specialties and discovered a high incidence of hand lesions in dermatologists compared to other clinical specialties (Lobraico, Schifano, and Brader 1988, 6–8). In 1989, a second survey was sent to 6,000 dermatologists who did not use lasers in their profession and the groups from both surveys were compared. Dermatologists in both groups reported lesions on the hands the most (89%) and those who did not use gloves during procedures reported twice as many cases of hand lesions as those who used gloves. The response may indicate hand lesions were acquired from direct contact and laser use in dermatologists did not lead to a noticeable increase in wart lesions (Lobraico, Schifano, and Brader 1989, 47). A follow up study by Gloster et al. (1995) sent surveys to 4,200 members of the American Society for Laser Medicine and the American Society of Dermatologic Surgery regarding their use of the CO₂ laser for treating warts and the safety precautions used during treatments. Although the response rate was low (14%), the authors concluded that the prevalence of warts in dermatologists who performed wart ablation procedures was not affected by the absence of controls during medical
laser procedures and the prevalence of warts in dermatologists was not statistically different compared to the general population (Gloster and Roenigk 1995, 436–441).

1.5.2 Health Effects in Animal Studies

Health effects from LGAC exposure have been reported in animal studies. Two studies vaporized porcine skin and exposed laboratory animals to the resultant LGAC; one used an Nd:YAG laser in CW mode at 15–20 W (Wenig et al. 1993, 242–245), while a second study used a CO2 laser at 20 W with a 1.5–2 mm beam diameter (Bagghish and Elbakry 1987, 1260–1265). Both studies reported pathological changes in laboratory rats similar to interstitial pneumonia, bronchiolitis, and emphysema (Wenig et al. 1993, 242–245; Baggish and Elbakry 1987, 1260–1265), conditions similar to human health effects studies from long-term inhalation of fine particulate matter PM from asbestos, tobacco, and talc (Baggish and Elbakry 1987, 1260–1265; Weiss, Dorow, and Felix 1983, 338–345; Morais et al. 1982, 21–31; Churg and Wiggs 1985, 364–372; Hoidal and Niewoehner 1982, 548–552). Both studies suggested that rats became sluggish and stopped active movement immediately after unfiltered LGAC exposure and resumed activity within 2 minutes post-exposure (Wenig et al. 1993, 242–245; Baggish and Elbakry 1987, 1260–1265). A subsequent analysis of laboratory rats exposed to 16 minutes of LGAC plume per day for two weeks showed severe pulmonary pathological changes (Bagghish and Elbakry 1987, 1260–1265); simple filtration (particles >0.5 µm) led to a decrease in pathological changes and high filtration (particles >0.1 µm) of the LGAC did not trigger any pathological change. Changes in behavior, including activity and feeding habits did not change with LGAC exposure (Bagghish, Baltoyannis, and Sze 1988,
Another study exposed laboratory rats to LGAC for 34 hours over the course of six months, followed by necroscopy. One rat developed angiosarcoma, none developed HPV infections, and no viral infections or DNA hybridizations were observed in the lungs of the animals (Nahhas et al. 1991, 259–262). A similar study exposed eleven sheep to the LGAC produced from the Nd:YAG laser irradiation of bronchial sheep tissue and found severe inflammation in the lungs after 10 and 30 minutes of exposure (Freitag et al. 1987, 283–288).

Animal studies have demonstrated LGAC exposure produces adverse health effects in rats and sheep but it is unclear how these results relate to human exposures. Only Nahhas et al. (1991) (Nahhas et al. 1991, 259–262) investigated exposures for longer than two weeks and all studies exposed animals to LGAC concentrations that were higher than levels experienced in clinical settings (Freitag et al. 1987, 283–288; Wenig et al. 1993, 242–245; Baggish and Elbakry 1987, 1260–1265; Nahhas et al. 1991, 259–262). The general results from the study indicate acute exposures lead to pulmonary inflammation and sluggishness in animals (Freitag et al. 1987, 283–288; Wenig et al. 1993, 242–245; Baggish and Elbakry 1987, 1260–1265). Long term exposures show an absence of health effects (Nahhas et al. 1991, 259–262), but results are dependent on one study with a limited number of subjects. Investigation into human health effects is nonexistent, only surveys have been administered and have not found any significant health outcomes (Lobraico, Schifano, and Brader 1988, 6–8; Gloster and Roenigk 1995, 436–441). To better understand possible adverse health outcomes related to LGAC exposure an investigation can be undertaken that studies short-term effects to the pulmonary
system by measuring peak expiratory flow (PEF). Long-term effects can be investigated by performing case-control studies that examine health outcome differences between groups.

1.6 Mitigation of Laser-Generated Air Contaminant in Clinical Settings

1.6.1 Importance of Control Strategies for Laser-Generated Air Contaminants

Mitigation of LGAC exposure is important because we believe the majority of particles in are in the alveolar hazard size-range (Kunachak and Sobhon 1998, 278–282). Ball et al. (2001) describes four issues associated with surgical smoke; odor, particle diameter, viability, and evacuation of the smoke during endoscopic procedures (Ball 2001, 125–142). Laser masks, local exhaust ventilation (LEV) and universal precautions were recommended control strategies; but it is unknown if these control strategies are sufficient for health protection. Some research has indicated that even with LEV use, viable material is ejected and can be found deposited at a distance (Wisniewski et al. 1990, 1117–1123).

1.6.2 Protection of Clinical Staff from Laser-Generated Air Contaminants

The National Institute for Occupational Safety and Health recommends a combination of LEV and general exhaust ventilation for control of the LGAC. The recommendation includes the use of a smoke evacuator with a capture velocity between 100 and 150 feet per minute and the use of HEPA filters that must be properly disposed of after each procedure (National Institute for Occupational Safety and Health (NIOSH) 1998). Universal precautions including the use of surgical masks and gloves are also recommended.
The American National Standards Institute recommends the use of a smoke evacuator or suction device placed in close proximity to the LGAC generation site. Filters are recommended when air is recirculation back into the room environment and should be disposed of properly; the use of universal precautions are suggested to deter transmission of infectious pathogens. However, ANSI does not believe the use of surgical masks is an effective tool for controlling LGAC exposure and is not recommended (American National Standards Institute (ANSI) 2005).

The Occupational Safety and Health Administration is the only US agency that can enforce occupational standards, but they do not have specific requirements relating to LGAC exposure and control. Instead, the agency depends on recommendations and standards from other organizations including ANSI and NIOSH and can cite or enforce those standards using the general duty clause.

Laser masks are respirators specifically marketed as PPE for LGAC protection and are effective in capturing particles larger than 5 µm, but may not offer any protection for particles less than 1 µm (Nezhat et al. 1987, 376–382; Lewin, Brauer, and Ostad 2011, 636–641); LGAC particle size has been reported between 0.3 and 1 µm (Alp et al. 2006, 1–5; Kunachak and Sobhon 1998, 278–282; Nezhat et al. 1987, 376–382; Freitag et al. 1987, 283–288). When reduction in particle counts between surgical and laser masks were compared using a Portacount Plus particle counter, Derrick et al. (2006), found laser masks were marginally better than surgical masks and both were significantly less effective compared to a filtering half mask (rating: FFP2) respirator (p=0.02) (Derrick et al. 2006, 278–281). Surgical masks were recommended by NIOSH and OSHA for use during medical laser procedures, but
they are designed to protect healthcare workers against blood borne pathogens in droplet form (Chen 1992; Chen et al. 1994, 65–74) and are ineffective at filtering surgical smoke (Chen 1992; Chen et al. 1994, 65–74; Chowdhury et al. 2011, 507–512), viruses and bacteria (Gatti et al., 781; Occupational Safety and Health Administration (OSHA), 1910:134 section1(a); Occupational Safety and Health Administration (OSHA), 1910.1030 section d(3)(I)). Particles in the alveolar hazard region are able to penetrate through the filter material (Kunachak and Sobhon 1998, 278–282) and enter through unsealed areas around the mask.

Local exhaust ventilation, including smoke evacuators and wall suction systems, are effective in reducing exposure to laser-generated surgical plumes at the lasing site (Bigony 2007, 1013–1024; Fisher 1987, 191–194). However one study found particles 100 to 200 µm in diameter were ejected from the lasing site and deposit on the medical laser operator’s face and equipment at distances of greater than 1 m, even when a smoke evacuator was used (Wisniewski et al. 1990, 1117–1123).

One of the main reasons for the lack of smoke evacuator use during medical laser procedures was the cost associated with the purchase and use of these devices. The cost for HEPA filters can be up to $130 and are single use, and charcoal filters range between $15 and $40 each. Evacuator systems can cost between $1200 and $5000 (Fitzgerald and Diekman 1992, 2, 22). Some healthcare workers have misconceptions about when LEV devices should be used. In one survey, clinical personnel involved in supporting different clinical specialties were asked about the use of LEV during clinical procedures. Healthcare workers used LEV more regularly
for laser procedures compared to electrocautery procedures; but use was variable,
ranging between 83% and 17% and was dependent on the procedure (Edwards and

1.7 Sampling Methodologies for Laser-Generated Air Contaminant Particles

For this study, we built, validated and used a laboratory-based emission
chamber to collect LGAC particles during the lasing of porcine tissue while varying
laser operational parameters and characterized the emission rate, shape and size of
the LGAC. This study was intended to help establish methods for the collection and
characterization of the particle fraction of LGAC and to demonstrate how the data
could be used to determine occupational exposures. Results from this study will be
used in future research that will expand the study by adding additional laser types
and levels in order to characterize as many devices, procedures and settings as
possible and developing an understanding of parameters that promote the
generations of LGAC particles and viable material.

We utilized two real time particle counters to measure a particle count
concentration of generated LGAC. The smallest particles were measured using a
condensations particle counter (CPC) that measures particles between 0.02 and 1
µm in diameter. These types of devices operate by supersaturating the sampled
aerosol with a fluid causing the particles to grow by condensation and counting the
particles using optics. While this method allows for counting of particles at ultrafine
size ranges, these devices have limitations related to the maximum particle
concentration they are able to measure (Mordas et al. 2005, 543–552) and this can
limit the environments in which the devices can be used to only locations that do
not contain high concentrations of particles. Larger particles between 0.3 and 10 µm were measured using a handheld optical particle counter that sizes particles using a light scattering technique. The optical counting technique is able to measure high counts of particles at a large size range but some studies have found the method has difficulty sizing irregularly shaped particles (Jonasz 1987, 137–142; Szymanski, Nagy, and Czitrovsky 2009, 918–929).
2. CHARACTERIZATION OF SIZE-SPECIFIC PARTICLE EMISSION RATES FOR A SIMULATED LASER SURGICAL PROCEDURE

2.1 Introduction

Lasers used for healthcare applications impact almost all clinical specialties, enabling shorter healing time of wounds, drier operative sites, and reduced trauma to surrounding tissue (Pierce et al. 2011, 1302–1309). While patient care may be improved, one potential occupational hazard of medical laser use is the production of LGAC, produced when cell membranes undergo pyrolysis and combustion, or rupture and release steam and cell contents (Alp et al. 2006, 1–5). The aerosol is composed mostly of water (95%), with the remaining 5% composed of gas phase contaminants and particulate matter that may contain viruses, bacteria, and blood and tissue particles (Gonzalez-Bayon, Gonzalez-Moreno, and Ortega-Perez 2006, 619) that may aerosolize, remain viable, and infect healthcare workers (Calero and Brusis 2003, 790–793; Hallmo and Naess 1991, 425–427). The smallest particles in the plume are produced from the uniform evaporation of liquids and are generally spherical in shape, composed of sodium, potassium, magnesium, calcium, and iron (Descoteaux et al. 1996, 152). Larger particles in the plume are produced mainly from the mechanical ejection of tissue fragments and contain carbon and oxygen (Weld et al. 2007, 347; Descoteaux et al. 1996, 152). It has been suggested that particle ejection (or “splatter”) is dependent in part on the power, wavelength, and PRF of the laser (Canestri 1999, 199–203).
Canestri et al. (1999) demonstrated that lower emission frequencies led to greater spatial separation of smoke, while higher power densities led to higher particle ejection speeds and deposition farther from the lasing area (Canestri 1999, 199–203).

Aerodynamic diameter standardizes particles into unit density spheres using their settling velocity (Hinds 1999) and is a determinant of location of deposition of the particles in the lungs (Shapiro et al. 1991, 123–123–134; Davies, Heyder, and Subba Ramu 1972, 591–591–600). It is believed that the transfer of viable cellular or viral material through the LGAC plume would be a function of particle size, most likely associated with larger particles. Ultrafine particles (<1 µm in Dₐ) are known to deposit deep in the lungs and may transfer directly into systemic circulation and affect secondary target organs (Kreyling et al. 2002, 1513–1530; Takenaka et al. 2001, 547–551; Ober dorster et al. 1995, 111–124; Nemmar et al. 2001, 1665–1668; Nemmar et al. 2002, 998–1004; Nemmar, Hoet, and Nemery 2006, A211-2; author reply A212-3; Schulz et al. 2005, 1–22).

The Occupational Safety and Health Administration estimates that more than 500,000 workers, including surgeons, nurses, anesthesiologists, and surgical technologists are exposed to laser or electrosurgical smoke (Occupational Safety and Health Administration (OSHA) 2008), yet little is known about the concentration and size fraction of the particles of LGAC, and previous studies focused nearly exclusively on CO₂ laser systems (Council on Scientific Affairs 1986, 900–907; Kunachak and Sobhon 1998, 278–282). The research that has been conducted has suggested adverse health outcomes from the inhalation of LGAC,
including pulmonary inflammatory response and viral infections (Freitag et al. 1987, 283–288; Baggish and Elbakry 1987, 1260–1265; Baggish, Baltoyannis, and Sze 1988, 248–253), and both OSHA and NIOSH have recognized the potential hazard that LGAC might pose to personnel who work within the vicinity of laser surgical procedures (Occupational Safety and Health Administration (OSHA) 2008; National Institute for Occupational Safety and Health (NIOSH) 1990).

Studies have reported variable size ranges and mean diameter of LGAC particles. Alp et al. (2006) found that particles generated using electrocautery devices are smaller (0.1 µm) than those produced during medical laser procedures (~0.3 µm) (Alp et al. 2006, 1–5). Kunachak et al. (1998) reported a larger mean laser-generated particle diameter by irradiating laryngeal papillomas with a CW CO₂ laser at 10 W and collecting particles using a 0.45 µm pore size Micropore filters. The filters were attached to the tip of a smoke evacuator hose and the evacuator was set to the maximum flow rate during the procedure. A SEM was used to count and estimate particle Dₐ, the authors measured particles ranging in size between 0.5 and 26 µm, with 80% being 0.8 µm Dₐ (Kunachak and Sobhon 1998, 278–282).

Freitag et al. (1987) performed experiments by irradiating 1 cm³ of sheep bronchial tissue using an Nd:YAG laser between 15 and 75 W. A seven-stage cascade impactor was used to collect the sample and each stage was weighed to determine the mass concentration. The mean aerodynamic diameter was determined to be 0.54 µm and mass per meter cubed of plume was 920 mg (Freitag et al. 1987, 283–288). Nezhat et al. (1987) collected laser plumes during 17 separate CO₂ laser treatments of endometriosis; the laser was used with a beam diameter of 0.5 mm and power
between 15 and 30 W. Samples were collected in sterilized plastic bags, that were subsequently sampled using a Marple personal cascade impactor for three minutes. The smallest stage of the impactor was designed to collect particles >0.2 µm. The authors determined a particle size distribution between 0.1 and 0.8 µm, with a median Da of 0.31 µm (Nezhat et al. 1987, 376-382).

Health effects from LGAC exposure have been reported in animal studies. Two studies—one using an Nd:YAG laser in continuous mode at 15 to 20 W (Wenig et al. 1993, 242–245), and the other using a CO2 laser at 20 W with a 1.5 to 2 mm beam diameter (Bagghish and Elbakry 1987, 1260–1265)—vaporized porcine skin and exposed laboratory animals to the resultant LGAC. Both studies reported pathological changes in laboratory rats similar to interstitial pneumonia, bronchiolitis, and emphysema (Wenig et al. 1993, 242–245; Baggish and Elbakry 1987, 1260–1265), conditions similar to those reported from long-term inhalation by humans of fine PM from asbestos, tobacco, and talc (Bagghish and Elbakry 1987, 1260–1265; Weiss, Dorow, and Felix 1983, 338–345; Morais et al. 1982, 21–31; Churg and Wiggs 1985, 364–372; Hoidal and Niewoehner 1982, 548–552). Both studies suggested that rats became sluggish and stopped active movement immediately after unfiltered LGAC exposure and resumed activity within 2 minutes post-exposure (Wenig et al. 1993, 242–245; Baggish and Elbakry 1987, 1260–1265). A subsequent analysis of laboratory rats exposed to 16 minutes of LGAC plume per day for two weeks showed severe pulmonary pathological changes similar to those reported in a previous study (Bagghish and Elbakry 1987, 1260–1265); simple filtration (> 0.5 µm) led to a decrease in pathological changes and high filtration
(>0.1 μm) of the LGAC did not trigger any pathological change. Changes in behavior, including activity and feeding habit did not change with LGAC exposure (Baggish, Baltoyannis, and Sze 1988, 248–253).

No studies have systematically examined the emission rate and particle size distribution of LGAC relative to laser operational parameters. Identifying parameters that influence particle size distribution will help operators determine appropriate laser operational parameter settings that minimize the emission of these particles. Findings may influence manufacturers to design systems that minimize particle emissions, and may be useful tools in estimating exposure concentrations and making risk judgments. To establish a method to systematically evaluate the effect of laser operational parameters on particle contaminant generation, we created a simulated surgical procedure using porcine tissue in an emission chamber to determine particle size distribution using two real-time particle counters with a porcine tissue model, and calculated emission rates for specific size fractions.

2.2 Methods

2.2.1 Emission Chamber

An emission chamber was designed, built, and validated as described in Lippert et al. (2013) (Lippert 2013, unpublished data); a technical drawing of the chamber is provided in Figure 1. Briefly, the chamber consisted of a square glass hood connected via a stainless steel transition section to an aluminum straight duct, and exhausted to a laboratory fume hood. Sampling probes for two real-time particle counters were spaced 2.5 cm from each other and were located more than
10 duct diameters (107 cm) downstream of the transition to allow the airstream to mix, stabilize, and reach laminar airflow. Both probes were positioned to face directly into the airstream; a 45-degree bend past the inlet of the probe allowed for a smooth transition out of the duct and reduced the amount of particles lost to impaction. Tygon tubing connected each probe to the respective particle counters. The AeroTrak® 8220 (TSI® Inc. Shorewood, Minnesota) measures particles between 0.3 and >10 μm at 6 size ranges (0.3–0.5, 0.5–1.0, 1.0–3.0, 3.0–5.0, 5.0–10, >10). The P-Trak® 8525 (TSI® Inc. Shorewood, Minnesota) ultrafine condensation particle counter measures particles between 0.02 and 1 μm. The chamber was validated by characterizing the velocity in the straight duct using a pitot tube attached to a micromanometer and calibrated against an orifice plate with two pressure taps to measure pressure differentials. Leak testing was performed by releasing a constant source of SF₆ gas into the chamber, and measuring the concentration using a MIRAN portable ambient air analyzer. Capture of the laser-generated smoke was visually confirmed using ventilation smoke tubes released in the emission chamber.

Isokinetic sampling, normally required when sampling particles in a gas, was not achieved due to the inability of the pumps on each particle counter to overcome the pressure difference necessary to match the airflow in the duct. Instead, the particle loss due to sampling in non-isokinetic conditions was calculated for each size fraction. A 12% loss was expected at the 10 μm micrometer size range, 6% at 5 μm and less than 2% for smaller particle size ranges. Results were not adjusted to reflect the particle loss since losses had little effect on the overall result.
Two medical laser systems were used: the Ultra MD™40 Laser System (max power = 40 W, $\lambda = 10,600$ nm, pulsed) (Laser Engineering Inc., Franklin, Tennessee) and the Medilas H 20-W Ho:YAG laser System (max power = 20 W, $\lambda = 2100$ nm, pulsed) (Dornier Medtech, Germany). The Ho:YAG laser is commonly used in removing stones from the urinary tract of patients, in the management of recurrent superficial bladder carcinomas, and in the treatment of urethral stricture (Syed et al. 2001, 625–627; Matsuoka et al. 2002, 968–972; Sun et al. 2001, 587–590). The CO₂ laser is used in the treatment of severe cases of rhinophyma, in surgical interventions of onychocryptosis, and in a number of dermatological applications, including skin resurfacing and tightening and in the treatment of severe cases of acne scars (Madan, Ferguson, and August 2009, 814–818; Lazzeri et al. 2012; Serour 2002, 509–512; Omi et al. 2011, 294–300; Collawn 2010, 526–529; Tanzi and Alster 2003, 80–84). We used porcine tissue in our laboratory simulation study because of its similarity to human skin, its use in previous LGAC studies (Plappert et al. 1999, 29–41; Stocker et al. 1998, 145–154), and its accessibility and low cost.
Figure 1. Emission chamber technical drawing.
2.2.2 Experimental Design

Three laser parameters were evaluated at settings that reflected the operational range for each device. Power (W), beam diameter (mm), and PRF (Hz) were tested at a high, low, and center level in order to try to determine which parameters influence emission rate. The range for beam diameter, PRF, and pulse duration were different for each device. The differences in manufacturer settings made the direct comparison of the lasers difficult. Instead, the lasers were tested separately from each other using levels that were optimum for each device.

The Ho:YAG laser allowed manipulation of the PRF between 1 and 20 Hz, the power between 1 and 20 W, and the beam diameter between 0.5 and 5 mm. The duration of each pulse was set by the manufacturer at 350 µs. Conversely, the CO₂ laser allowed manipulation of the PRF between 1 and 5 Hz, the power between 1 and 40 W, and the beam diameter between 0.5 and 2 mm. The pulse duration was adjustable between 0.1 and 0.9 seconds. The CO₂ laser was used in continuous shutter mode at the shortest pulse duration of 0.1 seconds, the mode which allows for continuous chopped power delivery of the energy at ≥0.1 seconds per pulse. However, a super-pulsed mode was also available on the device that delivers short pulse durations at high Hz, but the mode was not used since the PRF could not be controlled by the user and because the levels were significantly different than those available on the Ho:YAG laser.
To minimize the number of experiments, a $2^3$ full factorial design was used, consisting of eight unique experiments (Table II). Two replicate center points were utilized to test the linearity of relationship (Kutner et al. 2004; StatSoft). A test of curvature was conducted on the data and the standard deviation of the centerpoints was used as the experimental error of the design. Matched background samples were taken prior to each experimental sample.

### TABLE II
OPERATIONAL PARAMETERS USED FOR EXPERIMENTATION

<table>
<thead>
<tr>
<th>Unique operational parameter combinations</th>
<th>CO$_2$ laser parameter settings</th>
<th>Ho:YAG laser parameter settings</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Power (W)</td>
<td>Beam Diameter (mm)</td>
</tr>
<tr>
<td>1</td>
<td>5</td>
<td>0.5</td>
</tr>
<tr>
<td>2</td>
<td>5</td>
<td>2</td>
</tr>
<tr>
<td>3</td>
<td>5</td>
<td>0.5</td>
</tr>
<tr>
<td>4</td>
<td>5</td>
<td>2</td>
</tr>
<tr>
<td>5</td>
<td>12</td>
<td>0.5</td>
</tr>
<tr>
<td>6</td>
<td>12</td>
<td>2</td>
</tr>
<tr>
<td>7</td>
<td>12</td>
<td>0.5</td>
</tr>
<tr>
<td>8</td>
<td>12</td>
<td>2</td>
</tr>
<tr>
<td>Center point</td>
<td>8</td>
<td>1.25</td>
</tr>
</tbody>
</table>

Experimental runs were replicated three times to reduce the variability within each operational parameter combination. Experimental order of each experimental run was randomized and performed over the course of two days. An
ANOVA procedure was performed with SAS 9.2 (Cary, North Carolina) statistical software.

2.2.3 Simulated Procedure

At the beginning of each sampling day, the AeroTrak® and P-Trak® devices were zeroed using a HEPA Zero Filter. Background particle concentration in the laboratory was determined prior to each experimental run by sampling in the emission chamber with the AeroTrak® and P-Trak® devices for 10 minutes with the exhaust fan on and without lasing. Porcine skin, including underlying adipose was weighed and placed in the emission chamber.

The Ho:YAG laser was prepared by attaching the laser fiber to a fiber tip holder to control the beam and to set the angle of orientation during experimentation. A metal prong was attached to the fiber tip holder to regulate the beam diameter. The CO₂ laser was built with a movable arm and did not need a fiber tip holder to control the beam. Instead, a metal prong was attached to the laser arm to regulate the beam diameter, and the arm was moved across the tissue at the same rate as described in Lippert et al. 2013 (Lippert 2013, unpublished data).

Volumetric flow-rate in the straight duct was set to 3.3 m³/min to prevent saturation of the particle counters. Both particle counters began to log data 10 seconds after lasing had commenced and sampled for 10 minutes by logging a data point every second. The AeroTrak® particle counter produced an average particle count per meter cubed of air sampled for each size range. The P-Trak® produced a minimum, average, and maximum particle count per centimeter cubed of air sampled.
Size-specific particle emission rates were calculated using the particle concentration multiplied by the volumetric flow-rate to determine the number of particles generated per minute and an adjusted emission rate was calculated by subtracting the background sample from the experimental sample.

2.3 Result

2.3.1 Background Measurements

Background particle emission rates were greatest at the smallest particle size range, but high variability was present at every size range (Table III). At almost all of the size ranges, the standard deviation was larger than the average. Graphical representations of the background variance by time of sample indicate higher variability for samples taken in the afternoon compared to the morning. However, when background emission rate was graphed by time and day of sample, most of the variability was between day and emission rate rather than the time the sample was taken (Figure 2).
TABLE III
BACKGROUND EMISSION RATE (PARTICLES/ MIN) BY SIZE RANGE FOR BOTH SAMPLING DEVICES

<table>
<thead>
<tr>
<th>Size Range</th>
<th>AeroTrak® μm</th>
<th>P-Trak® μm</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.3–0.5</td>
<td>$6.9 \times 10^8$</td>
<td>$0.02–1.0$</td>
</tr>
<tr>
<td>0.5–1.0</td>
<td>$1.7 \times 10^8$</td>
<td>$3.0 \times 10^8$</td>
</tr>
<tr>
<td>1.0–3.0</td>
<td>$1.8 \times 10^7$</td>
<td>$3.5 \times 10^7$</td>
</tr>
<tr>
<td>3.0–5.0</td>
<td>$2.7 \times 10^6$</td>
<td>$6.2 \times 10^6$</td>
</tr>
<tr>
<td>5.0–10.0</td>
<td>$7.5 \times 10^5$</td>
<td>$1.7 \times 10^6$</td>
</tr>
<tr>
<td>&gt;10.0</td>
<td>$9.0 \times 10^4$</td>
<td>$1.9 \times 10^5$</td>
</tr>
</tbody>
</table>

Average and St. dev.

Figure 2. Particles emission rate between 0.3 and 0.5 μm Dₐ for background samples by day and time.

2.3.2 Emission Rate Results

In 43 of 377 (11%) samples the adjusted emission rate was negative, indicating that background emission rate was higher than the matched experimental sample. Results were modified by replacing negative adjusted emission rates with an insignificantly small value and analyzing the data using the
modified dataset. Unit normal probability plots were generated for each size range and all followed a multimodal distribution, suggesting a mixture of influence from difference sources. Since the distributions were unique at each size range this suggested different influential source characteristics for each range; however, simple transformations of data did not improve the results. Average emission rates by parameter settings are presented in Table IV.

### Table IV

<table>
<thead>
<tr>
<th>Power (W); Beam diameter (mm); PRF (HZ)</th>
<th>AeroTrak 0.3–0.5 µm</th>
<th>0.5–1.0 µm</th>
<th>1.0–3.0 µm</th>
<th>3.0–5.0 µm</th>
<th>5.0–10.0 µm</th>
<th>&gt;10.0 µm</th>
<th>P-Trak 0.02–1 µm</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ho:YAG Laser</td>
<td>5; 5; 5</td>
<td>1.5*10^7</td>
<td>4.1*10^6</td>
<td>7.1*10^5</td>
<td>7.2*10^4</td>
<td>1.5*10^4</td>
<td>-9*10^2</td>
</tr>
<tr>
<td></td>
<td>5; 5; 12</td>
<td>5.0*10^6</td>
<td>8.7*10^5</td>
<td>2.1*10^5</td>
<td>2.8*10^4</td>
<td>6.4*10^3</td>
<td>6.2*10^1</td>
</tr>
<tr>
<td></td>
<td>5; 1; 5</td>
<td>4.5*10^8</td>
<td>1.7*10^8</td>
<td>3.2*10^7</td>
<td>4.7*10^6</td>
<td>1.4*10^6</td>
<td>1.6*10^5</td>
</tr>
<tr>
<td></td>
<td>5; 1; 12</td>
<td>1.3*10^8</td>
<td>4.0*10^7</td>
<td>7.2*10^6</td>
<td>8.9*10^5</td>
<td>2.3*10^5</td>
<td>1.8*10^4</td>
</tr>
<tr>
<td></td>
<td>8; 3; 8</td>
<td>4.5*10^8</td>
<td>7.6*10^7</td>
<td>1.1*10^7</td>
<td>1.4*10^6</td>
<td>4.1*10^5</td>
<td>5.2*10^4</td>
</tr>
<tr>
<td></td>
<td>12; 5; 5</td>
<td>1.1*10^8</td>
<td>4.1*10^7</td>
<td>7.9*10^6</td>
<td>9.7*10^5</td>
<td>2.0*10^5</td>
<td>9.8*10^3</td>
</tr>
<tr>
<td></td>
<td>12; 5; 12</td>
<td>2.0*10^8</td>
<td>3.4*10^7</td>
<td>3.8*10^6</td>
<td>4.7*10^5</td>
<td>1.0*10^5</td>
<td>1.9*10^2</td>
</tr>
<tr>
<td></td>
<td>12; 1; 5</td>
<td>9.3*10^8</td>
<td>5.1*10^8</td>
<td>8.8*10^7</td>
<td>1.5*10^7</td>
<td>4.2*10^6</td>
<td>3.7*10^5</td>
</tr>
<tr>
<td></td>
<td>12; 1; 12</td>
<td>1.9*10^9</td>
<td>5.3*10^8</td>
<td>8.2*10^7</td>
<td>1.4*10^7</td>
<td>4.0*10^6</td>
<td>5.0*10^5</td>
</tr>
</tbody>
</table>

| CO₂ Laser                             | 5; 2; 1.2            | 1.3*10^9   | 1.7*10^8  | 6.6*10^6  | 4.7*10^5    | 1.3*10^5 | 6.3*10^3       | 9.5*10^10 |
|                                        | 5; 2; 5              | 1.3*10^9   | 2.7*10^8  | 5.8*10^6  | 9.7*10^5    | 2.5*10^5 | 2.4*10^4       | 1.5*10^11 |
|                                        | 5; 0.5; 1.2          | 2.2*10^6   | 2.2*10^5  | 2.6*10^4  | 5.1*10^3    | 5.8*10^3 | 5.5*10^3       | 9.9*10^8  |
|                                        | 5; 0.5; 0.5          | 3.8*10^6   | 3.5*10^5  | 4.9*10^4  | 5.1*10^2    | 1.2*10^3 | 4.8*10^1       | 1.9*10^10 |
|                                        | 8; 1.25; 3.2         | 3.5*10^8   | 4.6*10^7  | 1.1*10^6  | 7.7*10^4    | 1.5*10^4 | 6.3*10^2       | 1.8*10^11 |
|                                        | 12; 2; 1.2           | 2.2*10^9   | 7.0*10^8  | 2.7*10^7  | 3.3*10^6    | 7.7*10^5 | 8.7*10^4       | 2.3*10^11 |
|                                        | 12; 2; 5             | 2.0*10^8   | 8.2*10^7  | 7.1*10^6  | 1.2*10^7    | 3.0*10^6 | 3.7*10^5       | 7.9*10^11 |
|                                        | 12; 0.5; 1.2         | 2.3*10^7   | 3.0*10^6  | 2.9*10^5  | 2.3*10^4    | 2.3*10^3 | 1.4*10^3       | 2.5*10^10 |
|                                        | 12; 0.5; 5           | 1.3*10^9   | 1.6*10^8  | 3.6*10^6  | 2.6*10^5    | 5.2*10^4 | 2.2*10^2       | 3.6*10^11 |
Three-factor ANOVA without interaction terms was used to test effects of the three operational parameters as presented in Table V. Interaction terms were omitted from our analysis due to the small sample size. The F-test p-value for each operational parameter was used to indicate if statistically significant level means were present. The percent of the variation each operational parameter explains is also presented in Table V. In the Ho:YAG laser, an increase in power led to a statistically significant increase in particle emission rates for all size ranges except the smallest (0.02 to 1 µm). For the smallest, an increase in power led to a decrease in emission rate. A smaller beam diameter led to a statistically significant increase in emission rate at every size range, while PRF was only statistically significant at the smallest size range and decreased when pulses per second were increased. At all size ranges, beam diameter was the most influential variable in the emission of laser-generated particles, explaining the most variability in the results, followed by power. The influence of each variable increased as the particle size decreased.

Conversely, in the CO₂ laser an increase in power led to a statistically significant increase in emission rates between 0.3 and 3 µm and a decrease in emission rate at the smallest size range (0.02 to 1 µm), while a smaller beam diameter led to increased emission rates for particle in the smallest size range, between 0.5 and 1 µm, and 3 to 10 µm. A decrease was seen for particles from 0.3 to 0.5 µm, and between 1 and 3 µm. The PRF was statistically significant only at the smallest size range and decreased when pulses per second were increased. Beam diameter was the most influential variable in the emission of laser-generated particles followed by power at all size ranges except at the smallest range (0.02–1
µm), where power was the most influential variable followed by PRF. The operational parameters explained an increasing percent of the variability as the size range decreased.

### Table V

**Average Adjusted Emission Rate by Parameter Level and by Particle Size Range**

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Ho:YAG Laser (n=29)</th>
<th>AeroTrak®</th>
<th>P-Trak®</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0.3–0.5 µm</td>
<td>0.5–1.0 µm</td>
<td>1.0–3.0 µm</td>
</tr>
<tr>
<td>Power (W)</td>
<td>High (12)</td>
<td>2.5*10^8</td>
<td>3.6*10^7</td>
</tr>
<tr>
<td></td>
<td>Low (5)</td>
<td>6.7*10^7</td>
<td>2.2*10^6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>p&lt;0.002</td>
<td>p=0.009</td>
</tr>
<tr>
<td>% var. explained</td>
<td>20.7%</td>
<td>16.6%</td>
<td>13.0%</td>
</tr>
<tr>
<td>Beam Diameter (mm)</td>
<td>High (1)</td>
<td>2.6*10^8</td>
<td>4.5*10^7</td>
</tr>
<tr>
<td></td>
<td>Low (5)</td>
<td>5.1*10^7</td>
<td>2.0*10^6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>p=0.001</td>
<td>p=0.002</td>
</tr>
<tr>
<td>% var. explained</td>
<td>32.8%</td>
<td>31.1%</td>
<td>27.0%</td>
</tr>
<tr>
<td>PRF (Hz)</td>
<td>High (12)</td>
<td>1.8*10^8</td>
<td>3.9*10^6</td>
</tr>
<tr>
<td></td>
<td>Low (5)</td>
<td>1.3*10^8</td>
<td>2.6*10^6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>p=0.224</td>
<td>p=0.896</td>
</tr>
<tr>
<td>% var. explained</td>
<td>2.6%</td>
<td>0.0%</td>
<td>0.3%</td>
</tr>
<tr>
<td>CO2 Laser (n=25)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Power (W)</td>
<td>High (12)</td>
<td>4.4*10^8</td>
<td>7.2*10^6</td>
</tr>
<tr>
<td></td>
<td>Low (5)</td>
<td>2.2*10^8</td>
<td>2.1*10^6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>p=0.033</td>
<td>p=0.025</td>
</tr>
<tr>
<td>% var. explained</td>
<td>11.6%</td>
<td>14.4%</td>
<td>16.0%</td>
</tr>
<tr>
<td>Beam Diameter (mm)</td>
<td>High (0.5)</td>
<td>1.2*10^8</td>
<td>9.2*10^6</td>
</tr>
<tr>
<td></td>
<td>Low (2)</td>
<td>5.4*10^8</td>
<td>1.2*10^6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>p&lt;0.001</td>
<td>p=0.002</td>
</tr>
<tr>
<td>% var. explained</td>
<td>38.8%</td>
<td>31.4%</td>
<td>22.7%</td>
</tr>
<tr>
<td>PRF (Hz)</td>
<td>High (5)</td>
<td>3.8*10^8</td>
<td>1.2*10^6</td>
</tr>
<tr>
<td></td>
<td>Low (1.2)</td>
<td>2.8*10^8</td>
<td>1.0*10^6</td>
</tr>
<tr>
<td></td>
<td></td>
<td>p=0.383</td>
<td>p=0.442</td>
</tr>
<tr>
<td>% var. explained</td>
<td>1.8%</td>
<td>1.5%</td>
<td>4.8%</td>
</tr>
</tbody>
</table>
2.4 Discussion

Initial study design called for a $2^3$ fractional factorial study design to reduce the number of samples necessary in order to assess the impact of operational parameters on emission rate. After completing one full $2^3$ experimental design, only factors that were marginally significant would be included in the replicate run. However, experimental error in the design, measured by the standard deviation of the center points, was larger than the mean concentration for each size range, indicating a need for additional replicates of the full design. Two replicates were completed for each experimental run to reduce the error associated with the design. Examination of data collected from background and experimental runs helped to identify sources that contributed to the variability in the results. Background samples in particular were highly variable when examined as a single data set. At most of the size ranges, the standard deviation was greater than the mean, indicating a deviation of at least $\pm 50\%$. Variability was mostly between days and not within day. This result indicates that the laboratory environment contributed significantly to the variation in background concentration, and environmental controls, such as HEPA filtration of the incoming air, would be helpful in decreasing particle concentrations.

Many sources of variability affected the emission rate during the run sample. Porcine tissue used was purchased from a local butcher, and tissue thickness, texture, and color were different for each, similar to findings in the literature (Avon and Wood 2005, 30–39; Yager and Scott 1993, 531–533), which may have affected
the emission rate variability between tissue samples. Purchasing tissues from a scientific laboratory and performing additional samples for each operational parameter combination may help in reducing variability.

The delivery method for each laser also may have resulted in variability between samples. The Ho:YAG laser fiber had to be cut occasionally in order to maintain a consistent beam diameter, and some emission rate variability between fiber adjustments may have occurred. For the CO₂ laser, emission rate within samples may have been caused by adjustments in the lasing arm, which may have led to changes in output power.

In the Ho:YAG laser an increase in power, except at the smallest size range, and decrease in beam diameter led to higher emission rates. In the CO₂ laser an increase in power led to an increase in particle emissions between 0.3 and 3 µm, while particle emissions decreased at the smallest size range. The smaller beam diameter led to an increase in particle emissions at all size ranges except for particles >10µm. For both lasers an increase in PRF led to an increase in emission rate at the smallest size range.

The percent variability each operational parameter explains suggests beam diameter is the most influential variable in the emission of laser-generated particles, and explained between 25% and 60% of the variability in the Ho:YAG laser and between 12% and 38% of the variability in the CO₂ laser. In both lasers, the variability explained by the parameters increased as the particle size range decreased, indicating our operational parameters influence generation of the
smallest particles the most, and changes to operational parameters would have the largest effect on the emission of the smallest particle size ranges.

A prior study imaged the laser-generated plume from the lasing of polymethylmethacrylate (PMMA) with a CO₂ laser using a thermo-camera. Polymethylmethacrylate was used because of its uniform platform and because it is thermodynamically similar to human tissue. Measurements taken after lasing and from observing the plume using a thermo-camera found an increase in the generation of larger-sized particles when pulsed lasers were used, and possibly when energy per pulse was highest (Canestri 1999, 199–203). These operational parameter settings promote the mechanical ejection of particles. In our study, energy per pulse was highest at the lowest PRF for both lasers; however, our results indicate emission rate only increased for the smallest particles. This suggests particle emission rates may not be dependent on energy per pulse and increased energy per pulse may not increase mechanical ejection or material. Future research should continue to study the effect of power, beam diameter, and PRF and expand into other factors such as energy per pulse and beam duration that may affect generation of particles.

Steps can be taken to reduce much of the variance in the data. Automating the lasing of the tissue will allow for consistent skin-laser contact time at the same beam diameter. Utilizing particle counters intended for scientific data collection rather than environmental sampling can reduce uncertainty in the data and provide more flexibility during sampling, such as the ability to vary the flow rate of the device. Improving the air quality in the laboratory by placing HEPA air filters on the
air supply vents and sealing areas around the doorway will help decrease the fluctuations in background particle concentrations. Finally, increasing the duration of each sample and the number of samples at each operational parameter level can improve the variability between and within each sample.

During surgical procedures, clinical personnel must consider a number of human characteristics before deciding on the appropriate laser settings, including the amount of circulating blood, the specific heat, and thermal conductivity of the tissue (Pierce et al. 2011, 447–466). These human characteristics have a significant effect on the amount of energy that is absorbed by the skin, and may also affect the amount of LGAC generated. The operational parameter settings utilized during experimentation were chosen to study a range of settings for differences related to particle emissions.

Differences in delivery of the laser energy, maximum power, and PRF available for each laser did not allow for a direct comparison. The Ho:YAG laser used a laser fiber for delivery, which provided maximum flexibility in use, but also limited the effect at increased distance from the tissue; with increased use, the shape of the beam changed. The CO₂ laser used optics in a multi-hinge moveable arm to deliver the laser energy, dramatically reducing the divergence of the beam with increased distance, and created better control of beam diameter.

The PRF and the power were affected by the pulse duration. When power (W) is held constant, shorter pulse duration would increase the amount of energy (Joules) output by the laser. Higher energy per pulse is believed to increase the shockwave effect on tissue, increasing the amount of mechanically ejected material.
In our study, the Ho:YAG laser did not allow for manipulation of the pulse duration, it was set by the manufacturer to 350 µs. The CO₂ laser did allow for limited manipulation, but not to the level desired, the shortest pulse duration available in the continuous setting was 0.1 seconds, more than 280 times longer than for the Ho:YAG laser. This difference in pulse duration could have affected the type and amount of particles being generated by both laser devices.

Once lasing was initiated and particle generation commenced, particles had to travel from the site of lasing through the chamber and into the particle counters situated 100 cm past the transition of the duct. To minimize losses from the face of the chamber and impaction of surfaces, the airflow in the chamber was set to a 3.3m³/min. The necessary airflow made sampling at isokinetic conditions impossible, but calculations determined the losses in small particle ranges to be negligible. Even with such high airflow, the P-Trak® condensation particle counter became saturated during experimental runs with high particle generation. In most cases, an assessment of the data log determined the spike to be short, usually one or two data points in a 600 data point set. Saturation of the particle counter may have led to under reporting of the mean concentration.

2.5 Conclusions

The goal of this study was to establish a method of identifying laser operational parameters that influenced the generation of particles. Results indicate a need to further refine the collection method by reducing the variability present in the study design. We were able to demonstrate the importance of all three factors in the generation of particles during the lasing procedure.
2.6 Acknowledgements

This research and researchers were partially funded by NIOSH grant program # T42/OH008672. The researchers would like to thank the individuals who contributed and graciously volunteered their time to make this paper possible, including Sal Cali, Dr. Anders Abelmann, Dr. John Breskey, Dr. Chiping Nieh, Dr. Nurtan A. Esmen and Dr. Steven Lacey.
3. MICROSCOPY OF MEDICAL LASER-GENERATED PARTICLES FROM A SIMULATED SURGICAL PROCEDURE

3.1 Introduction

Every year more than 500,000 workers, including surgeons, nurses, anesthesiologists, and surgical technologists, are exposed to laser or electrosurgical smoke (Occupational Safety and Health Administration (OSHA) 2008). The LGAC is produced from the interaction between a medical laser beam and the target tissue. The generation of LGAC is mostly associated with the use of class 3B and 4 laser systems (American National Standards Institute (ANSI) 2007), the most powerful laser systems. The smallest particles in the plume are produced from the uniform evaporation of liquids and are generally spherical in shape, and composed of sodium, potassium, magnesium, calcium and iron (Descoteaux et al. 1996, 152). Conversely, larger particles in the plume are produced mainly from the mechanical ejection of tissue fragments and comprise carbon and oxygen. It has been suggested that particle ejection and splatter is dependent on the power, wavelength, and PRF of the laser (Canestri 1999, 199–203).

Three studies have considered the shape and size distribution of particles by observing the collected LGAC under a SEM. Nezhat (1987) sampled during laparoscopic treatments for endometriosis that used a CO₂ laser at a beam diameter of 0.5 mm and power of 15 to 30 W. Samples were collected on six-stage cascade impactors using Mylar film filters.
Particles were viewed using a SEM and described as round and homogenous with physical diameters equivalent to aerodynamic diameter. The mass median diameter obtained by microbalance was 0.36 $\mu$m and particle size ranged from 0.1 to 0.8 $\mu$m (Nezhat et al. 1987, 376–382). Kunachak (1998) collected and irradiated laryngeal papillomas from patients using a CO$_2$ laser at 10 W. The goal of the study was to evaluate the effectiveness of paper and cotton surgical masks in stopping infiltration of laser particles. Sampling was done on a 0.45 $\mu$m pore size microfilter attached to a smoke evacuator. The filters were exposed directly to the plume, or the smoke was passed through a cotton or paper surgical mask prior to collection on the microfilter. Five random areas from each filter were chosen for analysis using a SEM; particle size in the filters ranged between 0.5 and 27 $\mu$m for which 70% were 0.8 $\mu$m. The average particle density was 6 particles/mm$^2$. The authors conclude that the masks were ineffective at stopping particles from passing through, and may not protect against laser generate particles (Kunachak and Sobhon 1998, 278–282).

Taravella et al. collected particles by coupling a 25 mm methylcellulose filter to the inlet of a smoke evacuator during excimer laser ablation of the corneal stroma. The laser was operated at a fluence of 160 mJ/cm$^2$ using a 6 mm beam diameter and pulse frequency of 6 Hz. After sampling, the filters were removed and placed on covered petri dishes containing desiccant crystals for five days. Prior to SEM analysis, filters were coated with gold and two areas were chosen near the center of each filter for viewing; pictures were digitized and analyzed using a computer program. The mean particle diameter was 0.22 $\mu$m$\pm$ 0.056 (Taravella et al. 2001, 604–607).
Two other studies used energy-based cautery devices and collected the resultant plume for SEM analysis. Weld et al. (2007) studied particles generated from incisions of porcine muscle tissue using four different energy-based instruments. Particles were collected on electron microscope grids by depositing them using an electrical field. A SEM at unknown settings was used to observe the shape and diameter of the particles (Weld et al. 2007, 347). Descoteaux et al. (1996) sampled during electrocautery laparoscopic surgeries using a seven-stage cascade impactor to collect particles onto aluminum substrate foil and analyzed using a SEM. Sixty photographs were randomly taken and studied (Descoteaux et al. 1996, 152).

Viewing laser-generated particles under a SEM revealed that they are mostly in the fine and ultrafine range (Kunachak and Sobhon 1998, 278–282; Taravella et al. 2001, 604–607). None of the studies attempted to characterize the laser-generated particles when varying any parameter associated with the laser. Further, methods and materials used for collection, transport, and analysis of the plume differed between the studies, making it difficult to compare results.

The aim of this study was to qualitatively evaluate the range of particle sizes and shapes generated during laser tissue incision by collecting particles on filters and analyzing them using a SEM. This could inform whether particles were produced from combustion processes (i.e., cellular matter vaporization) or from mechanical processes (explosive ejection of cellular material), and if a quantitative assessment of particle size and shape may be undertaken in future studies. Laser parameter settings were varied between samples as a preliminary exploration of influence of operational parameters on particle generation.
3.2 Methods

Two lasers were used: the Ultra MDTM 40 CO2 Laser System (max. power = 40 W, wavelength = 10,600 nm, pulsed) (Laser Engineering, Franklin, Tennessee) and the Medilas H 20-W Ho:YAG laser (max. power = 20 W, wavelength = 2100 nm, pulsed) (Dornier Medtech, Germany). Lasing was performed on porcine tissue samples. Particulate matter was collected on 37 mm SKC polycarbonate filters with a pore size of 0.8 µm. Polycarbonate filters were used because they enable capture of ultrafine particles on the filter surface, and the surface facilitates particle viewing in SEM. The filters were secured in a closed-face cassette and the cassette was connected to an SKC Aircheck® 2000 sampling pump with a flow rate of 2.8 L/min for 20 seconds during the lasing procedure, allowing particles to adhere to the filter. The filter cassette was held 2 cm from the surface of the porcine skin in a similar fashion as described by Taravella et al. (2001, 604–607). Sample collection times were brief to prevent particle over-sampling and samples were collected consecutively on the same day. An exhaust fan was operated between samples to remove generated particles, and a background sample was collected in room air prior to the first experimental sample before preparing the filter for analysis in order to identify particles present in the laboratory environment. A laboratory blank sample was prepared, the sample was used by the microscopist to differentiate normal filter residue and structure from particles collected during the lasing procedure and to calibrate the SEM for optimal viewing. All filters were placed in petri dishes and in a desiccator for 36 hours before transport to a SEM laboratory. Subsequent analysis was performed at the University of Illinois at Chicago Research
Resources Center using a HITACHI S3000-N SEM. This instrument has a range of magnification from 15x to 200,000x and a resolution of 5 nm. One quarter of each filter was taped onto a large specimen holder for SEM viewing; no coating was added to the filters. Four sections (200 by 200 µm in size) on each filter were chosen by scanning the field and selecting areas with heavy concentrations of particles. The images were digitally captured at a resolution of 700x with an acceleration voltage of 20 kV using the backscatter electron imaging mode. Particles in the digitized image were counted and their size and shape described. Particle size was described as <2.5 µm, between 2.5 and 10 µm, or >10 µm, whereas particle shape was described as irregular or spherical. To observe differences in particle size or shape, samples were collected at two settings for both the CO₂ and Ho:YAG laser (Table VI).
### TABLE VI
LASER PARAMETER SETTINGS USED DURING LASING FOR EACH OF FILTER SAMPLES

<table>
<thead>
<tr>
<th>Laser</th>
<th>Sample</th>
<th>Parameter settings</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ho:YAG: 2100 nm</td>
<td>Sample1: Low setting</td>
<td>Power: 12W, Beam diameter: 0.5mm, PRF: 5Hz</td>
</tr>
<tr>
<td></td>
<td>Sample2: High setting</td>
<td>Power: 12W, Beam diameter: 0.5mm, PRF: 12Hz</td>
</tr>
<tr>
<td>CO₂: 10,600 nm</td>
<td>Sample3: Low setting</td>
<td>Power: 12W, Beam diameter: 1.0mm, PRF: 1.2Hz</td>
</tr>
<tr>
<td></td>
<td>Sample4: High setting</td>
<td>Power: 12W, Beam diameter: 1.0mm, PRF: 5Hz</td>
</tr>
</tbody>
</table>

### 3.3 Results

A background sample did not reveal any visible particles present on the filter surface (Figure 3). When operating at the high parameter settings (Figure 4A and 4B), the concentration of particles was greater for the CO₂ laser compared to the Ho:YAG laser. For the CO₂ laser, the particle size distribution varied from <1.0 to 20 µm. Thirty percent of the particles viewed were less than 2.5 µm, 60% were between 2.5 and 10 µm, and 10% were greater than 10 µm. The filter used in combination with the Ho:YAG laser collected a lower concentration of particles ranging in size from <1.0 to 10 µm. Particle shapes in the Ho:YAG sample were all spherical and smooth, in the CO₂ laser particles included spherical and irregular shapes. Most particles >10 µm were irregular in shape.
Figure 3. Background filter sample.
Figure 4. SEM photographs of both laser devices and settings. A.) Filter from high PRF settings using the Ho:YAG laser. B.) Filter from high PRF settings using the CO₂ laser. C.) Filter from low PRF settings using the Ho:YAG laser. D.) Filter from low PRF setting using the CO₂ laser.

Figure 4C and 4D are examples of images taken with both lasers when the PRF was set to 5 for the Ho:YAG laser and 1.2 for the CO₂ laser and while keeping all other factors at the high setting. The filter used in combination with the CO₂ laser
had the highest concentration of particles of all samples. Particles <2.5 \mu m in
diameter made up 75% of all particles with at least half being <1.0 \mu m. Size ranged
from <1.0 to >50 \mu m. Particles greater than 10 \mu m were irregular in shape.
The filter used in combination with the Ho:YAG laser at the low PRF setting also
showed an increase in concentration. Only 10% of particles were considered to be
<1.0 \mu m, with 90% between 2.5 and 10 \mu m. Fiber-like material was present
throughout the filter area, with strands varying in length from 5 \mu m to greater than
100 \mu m and a width of <1.0 to >5 \mu m. The range of particle size varied from <1.0 to
10 \mu m in diameter.

3.4 Discussion
This pilot study demonstrated a method to examine particle morphologies
and compare those morphologies between varying laser parameter settings. Some
irregularly shaped particles were believed to be aerosolized by mechanical ejection
as a product of the shockwave effect from each laser pulse. However, we could not
make a definitive assessment on most particles present on the filters, as only shape
and surface characteristic could be used to make such determination.

Samples taken from the Ho:YAG laser may represent the most appropriate
example of those produced from both mechanical and combustion processes. While
the sample collected at the high setting only produced spherical particles and at a
very low concentration, the low PRF setting produced a higher particle
concentration with a larger size range (<1.0 to 10 \mu m) and with many irregular
shapes. Some of the irregular particles may be a direct outcome of the reduction in
the PRF, which caused an increase in the energy per pulse. The increased energy may have produced a shockwave effect in the tissue (American National Standards Institute (ANSI) 2005) and subsequent mechanical ejection. Previous studies have shown the ability of some ejected material to be cellular and viral in nature, with the possibility of viability (Descoteaux et al. 1996, 152).

Samples taken from the CO₂ laser showed a range of particle size that was much larger than for the Ho:YAG laser (<1.0 to 50 µm) with a high concentration of particles <1.0 µm present at the low PRF settings. Although the resolution of the electron microscope was not able to closely investigate the shape and size of these particles, the smooth surface of the polycarbonate filter allowed us to view and count these particles. Particles <1.0 µm were visible on the sample used in conjunction with the Ho:YAG laser, but only when the PRF was low, and at a lower concentration than in either of the CO₂ laser samples. Increased levels of particles <1.0 µm may be an indication of high energy absorption by tissue and subsequent thermal vaporization (Yen 1997, 41). In samples collected from both lasers, some particles did not resemble a combustion product or a conglomerate of particles. These particles had cratered surfaces resembling gas bubbles and are products of foam formation created from the trapping of gases within solid or liquid media. Some were spherical in shape with uniformly spaced cratered surfaces, while others were irregularly shaped.

This pilot study refined and examined the collection and characterization of particle morphology from the lasing of porcine tissue. A single sample was taken at four different settings, and small filter areas from each sample were examined and
characterized. Studies of the plume using real-time particle counters have shown high variability in the particle generation rate over time (Lopez 2013, unpublished data). In order to attribute differences in particles shape, size, and concentration between factor settings, additional replicate samples need to be collected, and additional areas on each sample be examined. Expanding the collection to other parameter levels will also help in explaining factors that lead to differences in particle morphologies. Initial study design called for estimating particle concentration in a quantitative manner by randomly selecting areas on each filter and calculating particles per area. However, filter particle density was not evenly distributed and particle load was low in some samples, indicating a need to examine large areas on each filter and take many samples before a representative concentration could be determined. Adding a time component dependent on the expected emission rate outcome may increase the particle density at lower emission rate settings and decrease the area and number of samples necessary for a representative sample.

Experimentation for this study was done on porcine tissue since it has been extensively validated as a model for human skin (Dick and Scott 1992, 640–645). Although tissue used for our experimentation was not live, it is believed to be a reasonable surrogate for our purposes.

3.5 Conclusion

This study presents results that may demonstrate an ability to distinguish particle morphologies of LGAC plumes while varying factor settings during lasing of porcine tissue. The analysis of the results shows that additional samples would be
necessary to make a clear determination if differences in particle morphologies are related to differences in laser wavelength or other factor settings. Determination of mechanism of formation was difficult because we could only observe particle shape and size; however, some particles were characterized as forming from both mechanical and vaporization mechanisms. An expansion and refinement of these methods would be useful in determining how particle shapes differ when other laser parameters are varied, in improving our understanding of the effect particles may have on health, and in changes necessary in personal protective equipment in order to protect workers.

3.6 **Acknowledgements**

This research and researchers were partially funded by NIOSH grant program #T42/OH008672. The researchers would like to thank the individuals who contributed, and graciously volunteered their time to make this paper possible including Mr. Sal Cali, Dr. Anders Abelmann, and Dr. Alan Nicholls.
4. APPLICATION OF TWO-ZONE MODEL TO ESTIMATE MEDICAL LASER-GENERATED PARTICULATE MATTER EXPOSURES

4.1 Introduction


In a previous study, we measured particle emission rates during the lasing of porcine tissue in a simulated medical laser surgical procedure (Lopez 2013, unpublished data) using An Ultra MDTM40 Laser System (max power = 40 W, \( \lambda = 10,600 \) nm, pulsed)(Laser Engineering Inc., Franklin, Tennessee) to determine size-specific particle emission rates in an emission chamber (Lippert 2013, unpublished data).
Particle count concentrations (particles/m³) were obtained using two real-time particle counters at seven particle size ranges between 0.02 and >10 µm. The particle concentrations were converted to particle count emission rates by size range.

We utilized a two-zone model, which has been applied to solvent exposures (Nicas 1996, 542–550; Nicas, Plisko, and Spencer 2006, 284–291), and PM in a welding operation (Boelter et al. 2009, 298–306). The model is a method of accounting for imperfect air mixing within a room (Nicas 1996, 542–550). The location of contaminant generation is divided into a near-field and far-field zone, and concentration for each is modeled by calculating an inter-zone airflow (β), or the air exchange between the two zones, and assuming perfect air mixing within each zone. In our study, the near field contains the emission point source and the breathing zone of the laser operator and the far field comprises the remainder of the clinical suite. A two-zone model was preferred over a single source completely mixed-space model since it more accurately estimates contaminant concentrations and is a better estimate of occupant exposures (Nicas 1996, 542–550).

The interflow term β (m³/min) was described by Nicas (2000) and is derived from the estimated surface area (SA) of the geometrical shape and wind speed(s) of the near field (4.1) (NICAS 2000, 51-56).

\[
\beta = \frac{1}{2} \cdot SA \cdot S
\]  

(4.1)
Estimates for $\beta$ in the literature were reported in a number of studies, but none for indoor environments with high air volume exchange rates where $\beta$ is expected to be high. Baldwin et al. (1998) conducted wind speed surveys for 55 different work environments, including offices, warehouses, and light and heavy industry sites. A mean wind speed of 18 meters/min was reported; however, 85% of all measurements were below this value; a more useful estimate was believed to be the geometric mean of 3 m/min (Baldwin and Maynard 1998, 303–313). Boelter et al. (2009, 298–306) collected PM during welding activities in both indoor and outdoor locations, and estimated the interflow term directly by solving for $\beta$ using the near-field steady-state equation, arriving at $\beta$ values between 1 and 30 m$^3$/min. However, they assumed the plume generated from welding mixed instantaneously in the near field, which may overestimate $\beta$ (m$^3$/sec) since some of the plume escaped directly into the far field (Nicas et al. 2009, D69–71; author reply D71).

Cherrie et al. 1999 estimated the $\beta$ (m$^3$/sec) term using (4.2).

$$\beta = 0.0056 = \frac{Q}{3} * X^{3}$$

(4.2)

Where $Q$ is the convective heat release rate (W) and $X$ is the distance above the source of the plume (m). They utilized three $\beta$ (m$^3$/sec) terms for modeling their results of 3, 10 and 30 m$^3$/min in an assumed volume of 8 m$^3$.

The generation rate of PM was obtained from laboratory experiments conducted to characterize emission rates by systematically varying laser operational
parameter settings indicative of medical laser surgical procedures. Two emission rate values from our laboratory study were chosen for modeling, and are representative of two procedures described in the literature. Rhinophyma is a type of chronic connective tissue hypertrophy and is considered a severe form of rosacea, characterized by a bulbous nose (Madan, Ferguson, and August 2009, 814–818; Anonymous 2010). Treatment for rhinophyma involves the reshaping of the nose using a laser (Anonymous 2010). Two studies described the CO$_2$ laser procedure in continuous mode between 10 and 20 W with a defocused beam diameter between 2 and 3 mm (Madan, Ferguson, and August 2009, 814–818; Lomeo, McDonald, and Finneman 1997, 740) and described total procedure time to be 15 minutes (Lomeo, McDonald, and Finneman 1997, 740). A separate study stated that the vaporization of the major tissue bulk occurred between 7.5 and 10 W, with a beam diameter (D$_L$) of 0.1 mm, and final sculpting performed using a defocused beam (Lazzeri et al. 2012).

A second procedure that utilized a CO$_2$ laser was the treatment of onychocryptosis (in-grown toenail) (A.D.A.M Medical Encyclopedia - Pubmed Health). Laser treatment involves the vaporization of the keratin matrix and lateral nail fold. In this procedure, the laser may be used in continuous mode at a power between 4 and 5 W and a beam diameter between 1 and 2 mm (Serour 2002, 509–512; Ozawa et al. 2005, 302–305).

In this study we attempt to model an estimated range of occupational exposures to PM for healthcare workers involved in medical laser applications. We used laboratory derived emission rates from a previous study and theoretical room
conditions and applied a two-zone exposure model to obtain a range of estimated PM concentrations over 15 minutes.

4.2 Methods

4.2.1 Particle Emission Rates

Particle count based emission rates from two different laser operational parameter settings (#1: P: 12 W, Dₘ: 0.5 mm, PRF: 5 Hz; #2: P: 5 W, Dₘ: 2 mm, PRF: 5 Hz) using a Ultra MDTM40 CO₂ Laser System (Laser Engineering Inc., Franklin, Tennessee) were converted to mass-based emission rates by assuming that all particles were spherical in shape and had a uniform density of 1,000 kg/m³. The median value in each size range was used to calculate the mass of particles (Table VII) (Lopez 2013, unpublished data). Particles larger than 10 µm were not included since the manufacturer did not specify a maximum collection diameter. Overlap in collection diameters between particle counters were adjusted by subtracting the AeroTrak® (TSI, Shoreview, Minnesota) emission rates between 0.3 and 1 µm from emission rates collected using the P-Trak® (TSI, Shoreview, Minnesota) and the size range was adjusted accordingly for analysis.
### TABLE VII
MASS EMISSION RATE BY SIZE RANGE FOR TWO MEDICAL LASER OPERATIONAL PARAMETER SETTINGS

<table>
<thead>
<tr>
<th>Avg. size (size range)</th>
<th>P: 12W, DL: 0.5mm, PRF: 5Hz</th>
<th>P: 5W, DL: 2mm, PRF: 5Hz</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>(µg/min) (% total mass)</td>
<td>(µg/min) (% total mass)</td>
</tr>
<tr>
<td>0.16 µm (0.02–0.3)</td>
<td>766.0 (86.9%)</td>
<td>328.1 (60.2%)</td>
</tr>
<tr>
<td>0.4 µm (0.3–0.5)</td>
<td>45.0 (5.1%)</td>
<td>45.0 (8.3%)</td>
</tr>
<tr>
<td>0.75 µm (0.5–1.0)</td>
<td>35.1 (4.0%)</td>
<td>60.4 (11.1%)</td>
</tr>
<tr>
<td>2 µm (1.0–3.0)</td>
<td>15.1 (1.7%)</td>
<td>24.4 (4.5%)</td>
</tr>
<tr>
<td>4 µm (3.0–5.0)</td>
<td>8.8 (1.0%)</td>
<td>32.5 (6.0%)</td>
</tr>
<tr>
<td>7.5 µm (5.0–10.0)</td>
<td>11.4 (1.3%)</td>
<td>54.5 (10.0%)</td>
</tr>
<tr>
<td>Total</td>
<td>881.4 (100%)</td>
<td>545 (100%)</td>
</tr>
</tbody>
</table>

#### 4.2.2 Two-Zone Model

The two-zone model was described by Boelter et al. (2009) using first-order differential equations as presented in (4.3) and (4.4).

\[
V_n DC_N = GD_T + \beta D_T - \beta C_F D_T - \beta C_N D_T \tag{4.3}
\]

\[
V_F DC_F = \beta C_N D_T - [\beta + Q] C_F D_T \tag{4.4}
\]

Where:

- \( V_N \) and \( V_F \) are the volumes of the near (N) and far field (F).
- \( C_N \) and \( C_F \) are the particle mass concentrations in mg/m³.
- \( G \) is the mass emission rate in mg/min.
- \( \beta \) is the airflow rate between the two fields in m³/min.
- \( Q \) is the supplied/exhaust net airflow rate in the room in m³/min.
- \( T \) is time in minutes (min).
We used $\beta$ values similar to those used by Cherrie (1999) (Cherrie 1999, 539–546). It was assumed the laser operator worked near the point source of the plume. The maximum distance between the operator and the point source would be 2 m, so we estimated the volume in the near field ($V_{NF}$) to be 8 m$^3$ (Nicas and Neuhaus 2008, 599–608). We utilized two $\beta$ terms (9 and 30 m$^3$/min) to simulate a low and high near-zone interflow.

There was concern that the largest particles would behave differently over time in the modeling scenario since these particles settle at a faster rate than smaller particles. The settling velocity ($V_{ts}$) was calculated in still air for 10 µm ($d_a$) particles assuming uniform density ($P_p$) using (4.5) (Hinds 1999); a settling velocity of 0.003 m/s was obtained. Assuming a well-mixed room, with point source 1.5 m above the floor, it would take approximately 9 minutes for particles of 10 µm to settle.

$$V_{TS} = P_p d_a^2 \frac{G}{18 \eta} \quad (4.5)$$

The largest particles collected were more dense than the smallest particles we sampled (<1 µm). However, in our results the largest particles accounted for a small percentage of the overall mass, and it was assumed they would most likely not affect our modeled results.

Minimum ventilation requirements and recommended room areas, as specified by American Institute of Architects (AIA) in collaboration with the US
Department of Health and Human Services, were utilized to construct theoretical settings (American Institute of Architects (AIA) 2006). Two theoretical room volumes (170 m$^3$ and 22.5 m$^3$) were chosen to simulate a range of operating or procedure room conditions in which a clinical procedure may take place. Both rooms were modeled with 3 m ceilings, and the simulated OR was given an area of 55.75 m$^2$ (600 sq. ft.). The simulated treatment room was given a minimum room size of 7.33 m$^2$, typical of an office room, which may be used for minor treatments. The AIA requires ORs to maintain a minimum of 15 air changes per hour (ACH), while procedure rooms need a minimum of 12 ACH, and 6 ACH are required in treatment rooms. In order to account for the different ventilation rates, each room was modeled using a high and low (6 and 15 ACH) air exchange rate using two operational parameter settings similar to levels used in medical laser procedures identified in the literature.

4.3 Results

4.3.1 Results for Simulated Operating Room

Near-field steady-state concentrations varied between 0.03 and 0.15 mg/m$^3$ and far-field concentrations varied between 0.01 and 0.08 mg/m$^3$ (Table VIII). Concentration ratios between the near field and far field ranged between 10:1 and 1:1 and the largest ratio was modeled when $\beta = 9$ m$^3$/min and $G = 0.881$ mg/min. The smallest ratio was modeled when $\beta = 30$ m$^3$/min and $G = 0.545$ mg/min. Under all conditions, a decrease in the interflow term led to an increase in concentration in the near field, but had no effect on concentration at the far field. Steady-state
conditions were nearly achieved within fifteen minutes with assumed continuous emission.
### TABLE VIII
TWO-ZONE MODEL RESULTS IN MG/M³ FOR SIMULATED OR AND TREATMENT ROOMS AT 5, 10, 15 MINUTES AND STEADY STATE (SS)

<table>
<thead>
<tr>
<th>Time after start of lasing</th>
<th>(ACH = 15)</th>
<th>(ACH = 6)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>β = 9 m³/min</td>
<td>β = 30 m³/min</td>
</tr>
<tr>
<td></td>
<td>C_{NF}</td>
<td>C_{FF}</td>
</tr>
<tr>
<td><strong>Simulated OR (170 m³)</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>G = 0.881 mg/min</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5min</td>
<td>0.11</td>
<td>0.01</td>
</tr>
<tr>
<td>10min</td>
<td>0.12</td>
<td>0.02</td>
</tr>
<tr>
<td>15min</td>
<td>0.12</td>
<td>0.02</td>
</tr>
<tr>
<td>SS</td>
<td>0.2</td>
<td>0.02</td>
</tr>
<tr>
<td>G = 0.545 mg/min</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5min</td>
<td>0.07</td>
<td>0.01</td>
</tr>
<tr>
<td>10min</td>
<td>0.07</td>
<td>0.01</td>
</tr>
<tr>
<td>15min</td>
<td>0.07</td>
<td>0.01</td>
</tr>
<tr>
<td>SS</td>
<td>0.07</td>
<td>0.01</td>
</tr>
<tr>
<td><strong>Simulated Treatment room (22.5 m³)</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>G = 0.881 mg/min</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5min</td>
<td>0.184</td>
<td>0.1</td>
</tr>
<tr>
<td>10min</td>
<td>0.23</td>
<td>0.14</td>
</tr>
<tr>
<td>15min</td>
<td>0.25</td>
<td>0.15</td>
</tr>
<tr>
<td>SS</td>
<td>0.26</td>
<td>0.16</td>
</tr>
<tr>
<td>G = 0.545 mg/min</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5min</td>
<td>0.11</td>
<td>0.06</td>
</tr>
<tr>
<td>10min</td>
<td>0.15</td>
<td>0.09</td>
</tr>
<tr>
<td>15min</td>
<td>0.16</td>
<td>0.1</td>
</tr>
<tr>
<td>SS</td>
<td>0.16</td>
<td>0.1</td>
</tr>
</tbody>
</table>
4.3.2 Results for Simulated Treatment Room

Near-field steady-state concentrations varied between 0.03 and 0.5 mg/m³ and far-field concentrations varied between 0.1 and 0.4 mg/m³ (Table VIII). Concentration ratios between the near field and far field ranged between 1:1.625 and 1:1 and the largest ratio was modeled when \( \beta = 9 \text{ m}^3/\text{min} \) and \( G = 0.881 \text{ mg/min} \). The smallest ratio was modeled when \( \beta = 30 \text{ m}^3/\text{min} \) and \( G = 0.881 \text{ mg/min} \). The steady-state concentration was greater at every setting compared to the simulated OR and were reached within fifteen minutes when ACH was high, but continued to rise at the lowest ACH. Maximum particle concentration occurred at low air ACH, high particle generation rate, and when \( \beta \) was lowest.

4.4 Discussion

In our modeled results the highest estimated steady-state concentration of 0.5 and 0.4 mg/m³, respectively, occurred in the treatment room when the air exchange rate and interflow were lowest (ACH = 6, \( \beta = 9 \text{ m}^3/\text{min} \)). The lowest concentration of 0.03 and 0.01 mg/m³, respectively, were estimated in the OR when the air exchange rate and interflow were highest (ACH = 15, \( \beta = 30 \text{ m}^3/\text{min} \)). In most scenarios, the concentration reached steady state within fifteen minutes of commencement of the simulated modeled procedure. We expected it would take longer to reach steady-state conditions in the OR, since the volume of the far field is larger than that of the treatment room, but at the low mass concentration being modeled, a difference was not noticeable.

In a previous study Albrecht et al. (2005) had collected respirable particle concentrations in a hospital OR during the lasing of porcine tissue using a CO₂ laser
at 20 W and beam diameter between 0.6 and 1.2 mm, and measured concentrations in the near field ranging between 0.59 and 1.69 mg/m³, while concentrations in the far field were between 0.16 and 0.31 mg/m³. The room was well ventilated, and concentrations decreased by a factor of ten every five minutes after lasing had stopped (Albrecht, Washce, Muller 2005, 455). Tanpowpong and Koytong (2002) measured PM concentration in a hospital OR that was poorly ventilated. They used a portable dust monitor to obtain one-hour average concentrations for PM₁₅, PM₁₀, and PM₂.₅. The highest average PM concentrations were measured during the second hour of laser vaporization procedures (PM₁₅ = 306.3 µg/m³, PM₁₀ = 246 µg/m³, PM₂.₅ = 228 µg/m³) and were three times greater than OR background and 100 times greater than concentrations in an adjacent office (Tanpowpong and Koytong 2002, 53–57). Freitag et al. (1987) irradiated 1cm³ blocks of bronchial tissue for ten minutes in an enclosed chamber using an Nd:YAG laser at between 15 to 75 W and measured a PM concentration of 0.92 mg/liter of smoke (Freitag et al. 1987, 283–288). Our near-field estimates ranged between 0.03 and 0.5 mg/m³ and far-field estimates ranged between 0.01 and 0.4 mg/m³, near-field and far-field estimates are in the range reported by Albrecht et al. (2005) and Tanpowpong and Koytong (2002). Similarity in results indicates our model is a good predictor of PM concentration in a hospital OR.

The PM was modeled in different room sizes as the size and type of room used during laser procedures can vary. The most involved laser procedures occur in surgical ORs, which follow design standards to help reduce infection transmission. The AIA requires ORs to provide a minimum of 15 ACH, three of which must be
comprised of fresh air. Ninety-nine percent filtration of particles >0.3 \( \mu m \) is required in certain procedures. Newer ORs vary in size, but the recommended size is 55.75 \( m^2 \) (American Institute of Architects (AIA) 2006).

Short-duration outpatient procedures can be performed in an ambulatory surgical center or in dedicated procedure rooms. Such facilities also have standards similar to those described for ORs (American Institute of Architects (AIA) 2006). However, in some private practice facilities, it is likely that some laser procedures are performed in office settings, with minimal ventilation available. These facilities might rely more readily on other types of controls such as local exhaust ventilation. In these cases, based on our model results, concentration of PM may increase rapidly, decay slowly, and reach a steady state at high concentration levels.

In the last few years, many new outpatient procedures have been developed that continue to use some of the same common medical laser wavelengths, but new innovations may affect the particle generation rate. While our experiments utilized the CO\(_2\) laser in continuous shutter mode, other treatments use the laser in a scanning, super-pulsed, and/or fractionated mode. These methods are popular in dermatological procedures such as skin resurfacing or acne treatment. The method may involve a laser that scans along an area of skin, creating a dot pattern of irradiated tissue (Omi et al. 2011, 294–300). This method of treating acne scars, or skin resurfacing, promotes the rapid regeneration of epithelial tissue with minimal side effects and short downtime.

Future efforts will involve further refinement of modeled exposures, accounting for a range of laboratory-derived emission rates to reflect varied laser
operational parameters and clinical applications, and validation of modeled results with field study.

While the concentration of PM was low in our scenarios, it is important to note that the majority of the mass was comprised of particles in the ultrafine range. Currently, there do not exist any standards that regulate ultrafine particles, but OSHA does regulate exposures to inhalable and respirable particulates in the workplace, the 8-hour time-weighted average (TWA) permissible exposure limit (PEL) is 15 mg/m³ for inhalable concentrations and 5 mg/m³ for respirable. Our data indicate that the levels in healthcare settings as they relate to our modeled scenarios and procedures are far below the PELs, although other federal agencies believe the PEL for particulates does not reflect findings in the literature. Some state agencies have adopted the National Ambient Air Quality Standards (NAAQS) and recommend these for indoor air quality. They recommend concentrations of less than 0.150 mg/m³ for PM₁₀ and 0.035 mg/m³ for PM₂.₅, over a 24-hour period (Illinois Department of Public Health 2011). When compared to these values, our simulated concentration may exceed these recommendations for short periods of time. A different pressing concern in clinical settings is the ability of some airborne particles to carry or be viable material, including bacteria, viruses, and fungi, and to infect patients and clinical staff. In these cases, reducing the particle concentrations may provide an added benefit of reducing infections and other adverse health outcomes associated with the contraction and inhalation of this type of PM.
4.5 Conclusion

The goal of this study was to model an estimated range of occupational exposures to PM for healthcare workers involved in medical laser applications. Our results were similar to concentrations measured in limited field studies conducted in hospital ORs. However, technologies continue to be developed and applications of medical lasers continue to grow. We plan to investigate other models to estimate potential exposures and study new applications and their potential for LGAC emission.

4.6 Acknowledgments

This research and researchers were partially funded grant # T42/OH008672. The authors would like to thank those individuals who contributed their time and effort into preparing this paper for publications including Sal Cali and Dr. Anders Abelmann.
5. CONCLUSIONS

5.1 Experimental Design

This study demonstrated methods practical for the characterization of LGAC particles. An emission chamber was built and validated, the chamber performed well and provided a platform ideal for reproducible experimentation. Generated particles from the lasing of swine tissue were collected using two real-time particle counters at nine different parameter settings and, the design was replicated two additional times. Results were analyzed using ANOVA to determine influential parameters. Emission rates were applied to a two-zone model to simulate occupational exposures in clinical settings. A limited number of samples were also collected for SEM analysis and a determination on shape, diameter and mechanism of formation was made.

5.2 Overall Results

Emission rates of particles generated during the lasing of porcine tissue while varying three operational parameter between two levels were collected, and ANOVA was used to test if the difference between parameter level means were different from each other. A statistically significant result indicated mean emission rates were different between parameter levels. Beam diameter explained the most variation in the results, followed by power; PRF explained a small amount of the variability.

In experiments that utilized the Ho:YAG laser, power and beam diameter were statistically significant at all size ranges. An increase in power or a decrease in beam diameter led to an increased emission rate for size-ranges between 0.3 and >10 µm
Ultrafine particle (0.02–1 µm) emission rates also increased when beam diameter was decreased, but decreased when power was increased. The PRF parameter was statistically significant for ultrafine particle emissions only, and an increase in PRF led to a decrease in emission rate.

Mean emission rates from CO₂ laser experiments were more variable, and produced less statistically significant results compared to the Ho:YAG laser. An increase in power led to a higher emission rate of ultrafine particles and a lower emission rate for particles between 0.3 and 3 µm; emission rate of particles >3 µm were not statistically significant. An increase in beam diameter led to increased emission rates of particles between 0.02 and 1 µm, 0.5 to 1µm and 3 and 10 µm, and a decrease in emissions for particles 0.3– 0.5 µm and 1–3µm. The PRF parameter was only significant for ultrafine particles and an increase in PRF led to an increase in particle emissions.

Laboratory-derived emission rates and theoretical room conditions were applied to a two-zone exposure model to obtain a range of estimated PM concentrations over 15 minutes. Results were compared to two-field studies (Albrecht, Washce, Muller 2005, 455; Tanpowpong and Koytong 2002, 53–57) that collected PM in surgical suites during medical laser procedures. Our near-field estimates ranged between 0.03 and 0.5 mg/m³ and far-field estimates ranged between 0.01 and 0.4 mg/m³, near-field and far-field estimates are in the lower-end of the range reported by Albrecht et al. (2005) and Tanpowpong and Koytong (2002). Similarity in results indicates our model is a good predictor of PM concentration in a hospital OR. Results were within OSHA and ACGIH limits for
respirable and inhalable PM but, the standards do not account for biologically active material.

Filter samples collected particles with various shapes and diameters and some mechanisms of formation were identified. The Ho:YAG laser (Appendix Figure 7 and 8) at the high setting only produced spherical particles and at a very low concentration; the low PRF setting produced a higher particle concentration with a larger size range (<1.0 to 10 µm) and with many irregular shapes. The shape and surface characteristics of some of the irregular particles seen in the low PRF setting were different than particles collected on all of the other filter samples and it is unclear if the particles are products of tissue/laser interaction or from contamination of the filter during preparation.

Samples from the CO₂ laser (Appendix Figure 5 and 6) showed a range of particle size that was much larger than for the Ho:YAG laser (<1.0 to 50 µm) with a high concentration of particles <1.0 µm present at the low PRF settings similar to the trend observed in the Ho:YAG laser and to emission rate results. The morphological examination of filter samples demonstrated a method to examine particle morphologies and compare those morphologies between varying laser parameter settings. However, to make judgments using filter samples and to compare the results to emission rates, many filters need to be collected for each parameter setting and analyzed in a quantitative manner.

Researchers and federal agencies agree that exposure to LGAC has adverse health effect. They have referenced animal studies that have found short exposures to LGAC produces pulmonary inflammation, interstitial pneumonia, and
sluggishness. However, it is unclear how the effects detected in animals are related to occupational exposures experienced by clinical staff. In addition, some case studies have suggested the LGAC can remain viable after aerosolization and can infect personal. Risk assessments have mentioned the inability to quantify the exposure dose as a shortfall to determining the risk associated with performing medical laser procedures.

As a first step to determining occupational exposures, this research examined the LGAC particle emission rates, size, shape, and mechanism of formation while varying laser operational parameters to characterize the LGAC and to determine if an expansion to additional laser systems, wavelengths, and operational settings is helpful. The results suggest that beam diameter is the most influential operational parameter, that differences exist in the shape and size of particles between and within laser wavelengths and that the results can be applied to estimate exposure of clinical staff.

5.3 Problems Identified

The study was designed as a $2^3$ fractional factorial design to reduce the number of samples necessary to assess the impact of operational parameters on emission rate. However, experimental error in the design, measured by the standard deviation of the center points, was larger than the mean concentration for each size range, indicating a need for additional replicates of the full design. Two replicates were completed for each experimental run to reduce the error associated with the design. Assumptions of normality and equal error variance were checked using ODS graphics option in SAS 9.3. Normal probability plots were generated for each size
range by laser wavelength and equal error variance was checked within each parameter. Data were generated using both matched and daily average background samples to determine if a less variable background would improve the results, but both methods produced similar outcomes. Normal probability plots of the generated emission rate for each size range had multimodal distributions and the error variances were not equal. The use of regression analysis to estimate emission rates was not possible due to the violation of normality assumption; however, ANOVA analysis is considered robust even when data are not normally distributed and do not contain equal error variance. An increase in Type I error may occur when both assumptions are violated (Glass, Peckham, and Sanders 1972, 237–288).

Three-factor ANOVA without interaction-terms was used to test effects of the three operational parameters. Interaction terms were omitted from our analysis due to the small sample size. The F-test p-value for each operational parameter was used to indicate if statistically significant level means were present. Weighted least squares (WLS) and robust regression were applied to the data but were not used since they did not significantly improve the results.

Initial experimentation revealed ultrafine particle concentrations saturated the P-Trak® sampler and the events were reduced by using a high velocity in the chamber (425 m/min). However, the high velocity prevented sample collection using isokinetic conditions and the losses from sampling in non-isokinetic conditions (Appendix Table IX) was calculated using equations 6.1 to 6.3 presented in the Appendix. Results were not adjusted to reflect the error in sampling since they had no effect on the end result.
After increasing the flow in the chamber, some experiments continued to saturate the P-Trak® sampler and caused an underestimation of particle concentration for some samples. A data-log of each sampling event revealed most saturation events were short spikes of 2 to 5 seconds in a 600-second data set and an adjustment would have little effect on the overall sample average. Two samples saturated the P-Trak® sampler for more than 15% of data points but were not censored from the data since they had high emission rates and were helpful in determining parameters that promoted LGAC particle generation.

Saturation events can be eliminated by diluting the incoming air-stream with a known volume of filtered air using Y-splitters, and adjusting the measured particle concentration to account for the diluted airflow. The method will eliminate saturation events but will require additional equipment, testing, and data collection. A simpler solution would be to increase velocity in the emission chamber and calculate the sampling error.

A background sample was collected prior to each experimental sample and the LGAC particle emission rates were calculated by subtracting the background emission rate from the experimental sample emission rate. In 43 of 377 (11%) data points, LGAC particle emission rates were negative (Appendix Table X), most were collected at the larger diameter size-range and from the CO2 laser (34/43), and may indicate large particles were not readily generated using the CO2 laser or there may be interference that limited the particle counters from correctly sizing large diameter particles. Variability in background samples may also have interfered with the results, especially when generated emission rates were small.
Our findings were limited by the variability present between replicate samples and by the operational design of the real-time particle counters. Filtering the incoming airflow using a HEPA filter can reduce variability in background samples, purchasing laboratory grade porcine tissues that are most similar to human tissues will reduce variability between samples, and automating the lasing procedure will allow for consistent tissue to laser contact time and remove human variability. Implementing these changes may reduce the number of replicates necessary for a statistically significant result. Future experimentation should use laboratory grade particle counters with better precision and able to measure high particle concentrations in airstreams while taking measurements at short intervals and logging data-points for later viewing.

5.4 Future Research

This study developed methods capable of measuring and characterizing LGAC particle emission rates in a laboratory based surgical simulation. The methodologies can now be improved and the methods expanded to measure emission rates in various clinical procedures that utilize medical laser devices. Many new outpatient procedures continue to be developed that use some of the same common medical laser wavelengths, but new innovations may affect the particle generation rate. Additional laser parameters can be included to better explain the effect laser settings have on LGAC particle generation.

Future efforts will involve further refinement of modeled exposures, accounting for a range of laboratory-derived emission rates to reflect varied laser operational parameters and clinical applications, and validation of modeled results.
with field study. The results may be applicable in estimating exposures and in
developing risk assessments for medical personnel present during medical laser
procedures.

Beside exposure to high LGAC particle concentrations, the viability of the LGAC
has been a growing concern in clinical settings and one that requires further
attention. Emission rate and filter sample results can be used with viable collection
methods to determine the likelihood viable particles will be generated and the
parameter settings that promote the generation of viable material. From these
findings, clinical staff can be prepared by applying different types of controls to
safeguard patients and staff from the health hazard.

Variability present in both background and experimental samples made the
comparison of laser devices difficult. Methods to reduce the variability are
presented in section 2.4. An important finding that should be investigated in future
research is the relationship between PRF, power, and pulse duration. When power
is held constant, a shorter pulse duration increases the amount of energy (joules)
output by the laser. Higher energy per pulse is believed to increase the shockwave
effect on tissue, increasing the amount of mechanically ejected material. In our
study, the pulse duration of the Ho:YAG laser was set by the manufacturer to 350 µs.
In the CO₂ laser the pulse duration could be manipulated, but the shortest pulse
duration available in the continuous setting was 0.1 seconds, more than 280 times
longer than for the Ho:YAG laser. The difference in pulse duration could have
affected the type and amount of particles generated by the laser devices as the
energy per pulse was higher in the Ho:YAG laser.
5.5 **Conclusions**

Results demonstrate the methods employed in this study can be used to characterize LGAC particles generated during medical laser procedures and to estimate exposures to clinical staff. With new laser technologies being developed and applications of medical lasers continually growing, an expansion of this study into additional wavelengths, parameters, and levels will provide insight into clinical laser procedures and laser settings that influence high-LGAC generation and that produce viable LGAC. The results can be used to inform clinical personal on the risks associated with specific procedures and protective equipment can be tailored to fit the need of clinical staff.
Losses from sampling in non-isokinetic conditions were calculated using the experimental velocity (425 m/min) and equations from the Aerosol Technology 2nd edition book (Hinds 1999).

Since isokinetic conditions were not met, a correction was calculated for all particle size fractions (0.5, 1, 3, 5, and 10 µm). Stokes number was calculated using (6.1).

\[
\text{Stk} = \frac{\tau U_o}{d_s}
\]  

(6.1)

Where \( \tau \) is the relaxation time \( U_o \) being the flow in the emission chamber, and \( d_s \) being the probe diameter. The variable \( \tau \) was calculated using (6.2)

\[
\tau = \frac{P_p d^2 C_C}{18 \eta}
\]  

(6.2)

Where \( P_p \) is the particle density, \( d \) the particle diameter, \( C_C \) the Cunningham correction factor, and \( \eta \) being the viscosity. Once Stokes number was calculated for each particle size fraction, (6.3) was used to obtain the correction factors for the different size fractions.
\[
\frac{C}{C_0} = 1 + \left( \frac{U_0}{U} - 1 \right) \left( 1 - \frac{1}{1 + \left( 2 + 0.65 \cdot \frac{U}{U_0} \right) \cdot \text{stk}} \right)
\]  

(6.3)

Where \( U \) is the velocity in the sampling probe, \( U_0 \) the velocity in the straight duct, \( C \) the concentration in the probe, \( C_0 \) the concentration in the free stream and \( \text{stk} \) the stokes number calculated in (6.1).

<table>
<thead>
<tr>
<th>TABLE IX</th>
<th>CORRECTED LGAC PERCENT SAMPLED BY PROBES</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>AeroTrak®</td>
</tr>
<tr>
<td>0.5 µm</td>
<td>1 µm</td>
</tr>
<tr>
<td>100%</td>
<td>100%</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>TABLE X</th>
<th>PERCENT OF SAMPLES WITH NEGATIVE GENERATED EMISSION RATES</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>AeroTrak®</td>
</tr>
<tr>
<td>0.3–0.5 µm</td>
<td>3/54</td>
</tr>
<tr>
<td>0.5–1 µm</td>
<td>2/54</td>
</tr>
<tr>
<td>1–3 µm</td>
<td>6/54</td>
</tr>
<tr>
<td>3–5 µm</td>
<td>1/52</td>
</tr>
<tr>
<td>1 µm</td>
<td>5/54</td>
</tr>
<tr>
<td>5–10 µm</td>
<td>5/54</td>
</tr>
<tr>
<td>&gt;10 µm</td>
<td>2/54</td>
</tr>
<tr>
<td>1 µm</td>
<td>6/54</td>
</tr>
<tr>
<td>Total</td>
<td>43/377</td>
</tr>
</tbody>
</table>
Figure 5. SEM photographs of CO2 laser filter
Figure 6. SEM photographs of CO2 laser filter sample using low laser settings (12 W, 1.2 Hz, 2.5 mm).
Figure 7. SEM photographs of Ho:YAG laser filter sample using high laser settings (12 W, 12 Hz, 1 mm).
Figure 8. SEM photographs of Ho:YAG laser filter sample using low laser settings (12 W, 5 Hz, 1 mm).
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- B.S., Applied Mathematics, University of Illinois at Urbana/Champaign, Urbana, Illinois, 2007
- M.S., Public Health Science, University of Illinois at Chicago, Chicago, Illinois, 2010
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