Computer modeling of the operating room ventilation performance in connection with surgical site infection

B. Sajadi¹, M.H. Saidi²*, G. Ahmadi³

¹ School of Mechanical Engineering, College of Engineering, University of Tehran, Tehran, Iran, Email: bsajadi@ut.ac.ir, Tel: (+98) 21 6111 9944, Cell phone: (+98) 912 289 7702
² School of Mechanical Engineering, Sharif University of Technology, Tehran, Iran, Email: saman@sharif.edu, Tel: (+98) 21 6616 5522
³ Department of Mechanical and Aeronautical Engineering, Clarkson University, Potsdam, NY, Email: gahmadi@clarkson.edu, Tel: (+1) 315 268 2322

Abstract

The primary source of surgical site infection is deposition of flakes released from exposed skin of surgical staff or the patient on the exposed surgical wound. In this study, a computational model for simulating airflow and thermal conditions in an operating room is developed, and transport and deposition of particulate contaminants near the wound are analyzed. The results show formation of a thermal plume over the wound tissue, which is typically at higher temperature than surrounding. The thermal plume protects the wound from deposition of contaminants. In addition, the computational model predicts an optimum value for the inlet air velocity that is mainly to maintain protective shielding effect of the wound thermal plume. The effects of particle size, surgical lights characteristics, and presence of partitions on the optimum inlet air velocity are also studied. Based on the results, formation of thermal plume over surgical lamps may easily disturb the airflow and impresses the optimum inlet air velocity accordingly. Unfavorable obstruction effects of surgical lights can be reduced using well-designed luminaries. The present study provides a better understanding of
airflow pattern and transport process in operating rooms equipped with UCV systems and may find application in designing more effective ventilation strategies.

**Keywords**
Operating room (OR), Ultra-clean ventilation (UCV) system, Computational fluid dynamics (CFD), Surgical site infection (SSI)

1. Introduction
Surgical site infection (SSI) is one of the most common post-surgical complications. According to the Centers for Disease Control (CDC) reports, SSI is the third most common nosocomial infection averaging about 14-16% [1]. Furthermore, 2.6% of all surgeries suffer from SSI [2]. The risk of SSI significantly increases for more prolong deep surgeries such as joint replacement. The additional expenses due to SSI were estimated about $5000 in 1999 [2] costing the health care system in US about $3.5 billion per year [3].

It is well known that the most common pathogen causing SSI is Staphylococcus aureus, which is responsible for 48% of all cases [2]. Staphylococcus aureus is a spherical bacterium naturally found in normal human skin. Staphylococcus aureus infections have become more challenging in the recent decades due to the appearance of several antibiotic-persistent strands of the bacteria [3]. The primary source of Staphylococcus aureus is the flakes which are shed from the exposed skin of the surgical staff or the patient. Typically, these flakes are 25 μm in diameter and 3-5 μm in thickness. It was estimated that during a two- to four-hour surgery, 1.15 to 90 million flakes are released [4], and about 5-10% of the flakes are contaminated with the bacteria [5]. According to [6] a surgeon, bending over the patient, may be a source of as many as 1000 airborne particles per minute. Furthermore, distance of the surgical staff to the operating table affects the risk of wound contamination.
The risk of SSI depends on several factors, including patient susceptibility to infection, surgical staff practices, operating room cleanliness and the HVAC system [7]. Although health professionals have successfully used various techniques such as sterilizing surgical instruments, training the staff, using antibiotics and developing the minimum invasive surgery (MIS) methods to prevent the surgical wound from infection, SSI still occurs. Woods et al. [4] categorized the source of wound infection into (1) patient, (2) non-sterilized instruments, and (3) airborne bacteria. It is believed that most SSI cases are due to airborne bacterial particles, which in fact could be controlled using a properly designed ventilation system. To minimize the harmful microbiological agents responsible for the infection of surgical sites, cleanroom technology was applied to the ventilation of operating rooms in 60s, which led considerable reduction in the number of postoperative SSI [8]. Recent studies of Lidwell [9], Charnley [10] and Ferrazi [11] for joint replacement, hip replacement and cardiac surgeries, respectively, also showed approximately 50% improvement in the wound infection control using proper ventilation airflow pattern.

Classification of the operating room ventilation systems is similar to the cleanroom ones. Accordingly, the ventilation systems are categorized to (1) mixed or conventional and (2) unidirectional or laminar systems. Unidirectional systems, which are also known as ultra-clean ventilation (UCV) systems [12], provide a protected clean area over the surgical table to minimize the transmission of airborne bacteria from less clean area. The use of UCV systems has been recommended for surgical procedure which deals with subcutaneous tissues or internal organs [13]. While the downward unidirectional ventilation system provides the most effective UCV systems, it has some disadvantages, one major one is that the flow may be disturbed by upstream obstacles especially surgical lights [14]. As a result, some other configurations such as horizontal [15], diagonal [16], differential [17], temperature-controlled [18], local [19] and portable [20] ones have been introduced. Numerical studies of Sadrizadeh
et al. [21] showed that the proper ventilation scheme highly depends on internal obstacles and staff work practice.

Since early 90s, with improvement in the computational capacity of computers, computational fluid dynamics (CFD) techniques have become an important engineering design tool to study airflow and particle deposition in complex geometries [22-24]. Application of CFD for investigating indoor air quality in operating rooms was reported in some researches. Accordingly, the flow characteristics in an operating room depends on its geometry and door opening cycles [25], ambient temperature [26], supply inlets and return outlets configuration [27], inlet diffusers geometry [28], ventilation airflow rate [29], location of obstacles [30], surgeons posture [31], staff clothing [32] and their activity level [33]. The results of previous studies showed that the location of return outlets is not as important as the location of supply inlets [34, 35]. Memarzadeh and Manning [7] suggested that using mixed high-low level outlets may lead to better airflow distribution; however, the differences are not significant enough to make low- or high-level systems unreliable. Although supply inlets arrangement has a considerable effect on the airflow pattern and the indoor air quality (IAQ) in conventionally ventilated environments, it seems that there is no concern regarding this issue in a UCV system, provided that the unidirectional zone covers the operating site and the surgical team. In early nineties, Chen et al. [29] investigated airflow and contaminant dispersion in an operating room using Eulerian-Eulerian numerical approach. They concluded that increasing the ventilation airflow rate would lead to better air quality. The same conclusion was also reported by some others [36]. This intuitive result, however, is contrary to some experimental data that suggested sometimes ventilation airflow increment does not show any improvement or even may have negative effects on the wound infection rate [37, 38]. It was suggested that this inconsistency between the computer model and the experimental data is due to treating the particulate phase as a scalar which causes missing the
physical nature of particles that affects their transport and deposition [7]. As noted before, bacteria-carrying skin flakes are of approximately 10 μm in diameter. In this size, the particle inertia effects are important and Eulerian diffusion approach cannot reflect all aspects of the particles behavior. Computational modeling of particle dispersion and deposition in complex geometry passages were reported by Li et al. [39], Ahmadi and Smith [40], and Zhang and Ahmadi [41], among the others. Memarzadeh and his coworkers [42] conducted extensive numerical studies investigating the effect of operating room ventilation system on reducing the risk of infection in surgeries. Using an Eulerian-Lagrangian code, they concluded that the indoor air pattern is more important than the ventilation rate to prevent the wound from infection. They also found that for the laminar airflow ventilation systems, there is an optimum value of 0.15 m/s for the inlet air velocity beyond which the particle deposition on the wound increases due to the distortion of the protecting wound thermal plume [7]. This observation was also supported by Rui et al. [43] and Sajadi et al. [44]. That is while the ventilation airflow increment leads to less contaminant concentration over the operating room, it not necessarily decreases the risk of wound infection. Despite the fact that Memarzadeh and Manning’s results are the basis for ASHRAE recommendations [45], the optimum inlet air velocity is not quite unique, as other standards suggest different values [12, 46]. Previous studies also reported that the surgical lamps have considerable influence on the airflow distribution all over an operating room due to both generating a thermal plume and obstructing the airflow [30]. Based on the results, the lights location [47] and their geometry [48] may remarkably affect the deposition of infectious airborne particles over the wound and should be considered to design an effective ventilation system.

In this paper, the airflow distribution through an operating room and the resultant particle deposition over the wound are studied numerically. The effect of inlet air velocity and formation of a protecting thermal plume over the wound on the velocity distribution and
airborne particles dispersion in the operating room is evaluated. Specific attention is paid to investigate the effect of surgical luminaries on the wound infection risk. The results of the present study in providing a better understanding of the airflow pattern in the operating rooms and the effect on surgical lights on it may find applications for developing more effective ventilation strategies to reduced infections after surgeries.

2. Numerical Method

Figure 1 shows a schematic of the operating room model with a typical area of \(37 \, \text{m}^2\) [49]. The arrangement of operating staff, equipment and patient in the modeled room are selected in accord with DIN 4799 recommendations [50]. Due to a complex geometry of the model and room contents, unstructured tetrahedral mesh is used to discretize the computational domain. A mesh with \(1.8 \times 10^6\) cells is used for the computational analysis. The mesh density is varied with denser mesh in the ultra-clean area where the gradients are predicted to be higher. A grid refinement was performed and it was found that the mesh size used is sufficient to guarantee the independency of the results from the number of grids.

For an incompressible fluid with heat transfer, the governing equations are,

\[
\nabla (\rho \mathbf{V} \phi - \Gamma_\phi \nabla \phi) = S_\phi
\]

where \(\mathbf{V}\) is the velocity vector and the effective diffusion coefficient, \(\Gamma_\phi\), and the source term, \(S_\phi\), for different parameters, \(\phi\), are listed in Table 1.

To resolve the turbulence closure problem, the RNG k-\(\text{e}\) model [51] accompanied by the standard wall function [52] is implemented to extract the turbulent properties due to its simplicity, robustness, and relatively accurate results [53]. The governing equations are solved on a collocated grid using ANSYS FLUENT 12.1 commercial package [54]. Pressure-velocity coupling is established through SIMPLE algorithm [55] and all convection terms are
discretized using second order upwind scheme which was found to provide an acceptable accuracy in the indoor airflow modeling [56].

It is well known that Eulerian-Eulerian approach is not appropriate for analyzing the bacterial airborne particles motion due to their relatively large size [7]. Therefore, Lagrangian trajectory analysis is used. The corresponding particle equation of motion is given as [44],

\[
\frac{d\mathbf{u}_p}{dt} = \frac{1}{\tau} (\mathbf{u}_i - \mathbf{u}_p) + g_i + n_i(t)
\]  

(2)

where \(\mathbf{u}_p\) is the particle velocity, \(g_i\) is the acceleration of gravity, \(n_i\) is Brownian force per unit mass, and \(\tau\) is the particle relaxation time defined as,

\[
\tau = \frac{S d_p^2 C_c}{18 \nu}
\]

(3)

where \(d_p\) is the particle diameter, \(S\) is the particle-to-fluid density ratio and Cunningham slip correction factor, \(C_c\), is included to account for non-continuum effects. The Cunningham correction factor may be given as,

\[
C_c = 1 + Kn \left[ 1.257 + 0.4 \exp \left( -\frac{1.1}{Kn} \right) \right]
\]

(4)

where Knudsen number, \(Kn\), is,

\[
Kn = \frac{2\lambda}{d_p}
\]

(5)

where \(\lambda\) is the air mean free path which equals 68 nm at the normal atmospheric conditions. Other forces, such as Saffman lift force, which are small, are neglected.

The turbulence diffusion which strongly affects particle transport in the indoor airflows needs to be properly estimated. In this study, discrete random walk (DRW) model is used to simulate the instantaneous turbulent fluctuating velocity components. Accordingly, the fluctuation velocity is given as [18],
\[ u_i' = G \sqrt{u_i'^2} \]  \hspace{1cm} (6)

where \( G \) is a Gaussian random number and \( \sqrt{u_i'^2} \) is the root mean square (RMS) of the \( i \)-th fluctuating velocity component. For the \( k-\varepsilon \) turbulence models, the mean square of fluctuation velocity components are equal to \( 2/3 \ k \).

The random number \( G \) is updated using the eddy lifetime and the particle crossing time. The characteristic lifetime of turbulence eddies is,

\[ \tau_e = 2T_L \]  \hspace{1cm} (7)

where \( T_L \) is the turbulent Lagrangian time scale. For the \( k-\varepsilon \) turbulence models, \( T_L \) can be estimated as,

\[ T_L = 0.15 \frac{k}{\varepsilon} \]  \hspace{1cm} (8)

The particle crossing time, the time a particle needs to pass across an eddy, is defined as,

\[ \tau_c = -\tau \ln \left( 1 - \frac{L_e}{\tau \mu - \mu_p} \right) \]  \hspace{1cm} (9)

where \( L_e \) is the eddy length scale.

The particle is assumed to interact with one eddy over the smaller value of \( \tau_e \) and \( \tau_c \); therefore, the random number \( G \) is updated after the minimum of \( \tau_e \) and \( \tau_c \).

2-1. Boundary Conditions

For the UCV system, constant velocity and constant temperature (typically 300 K) inlet boundary conditions are used. At the outlets all gradients are assumed to be zero. On the walls the no slip velocity boundary condition is used. A constant temperature of 300 K is used for the human (staff and patient) skin [14], except at the surgical wound where the temperature is assumed to be 310 K which is the core body temperature [7]. It is very
common to use constant heat flux boundary condition for the operating lights based on their power consumption. However, most of the power used by lights is transmitted via radiation mechanism which is not normally considered in indoor airflow investigations. Such a simplification in the heat transfer mechanism modeling leads to over prediction of the luminary surface temperature and the resultant thermal plume. In this study, a constant surface temperature of 350 K was assumed for the surgical lights. This value is consistent with the earlier experimental studies reported by ADMECO AG [57]. The particles behavior on the boundaries is also important for particle trajectory. In this study, it is assumed that particles are released from the wound with zero initial velocity. They are trapped on walls and exit from the computational domain as reaching an outlet.

3. Results and Discussions

Memarzadeh and Manning [7] suggested that only particles released close to the wound may lead to its contamination. In this study, to examine the effect of inlet air velocity on the wound infection, particles are released uniformly from 0.01 m$^3$ volume over the wound and the percentage of particles deposited on the wound is evaluated. The particles density is assumed to be similar to water, as bioaerosols are mostly made of water. The performed statistical independency study showed that $10^5$ particles are necessary to ensure that the percentage of the deposited particles is independent of the number of released ones. Figure 2 shows the effect of inlet air velocity and particle size on the percentage of particles that are deposited on the wound. It is seen that particles deposition is almost independent of the particle size in the range of 0.1 - 1 μm and increases rapidly for the larger ones mainly due to gravity. Participation of the deposition mechanisms, including gravity and impaction, in deposition of airborne particles on the wound is also presented in Figure 2. Due to a relatively large size of the studied particles, the effect of Brownian motion is almost negligible and the
main deposition mechanisms include only impaction and gravity force; the effect of the latter one vanishes rapidly as the particle size goes under 1 μm. Deposition of particles due to impaction depends not only on the particles concentration over the wound, but also on the near wound distribution of airflow. One interesting feature of the variation of deposition curves with velocity, shown in Figure 2 for different particle sizes, is that they all show an optimum value for the inlet air velocity which minimizes particles deposition. Furthermore, the optimum velocity is roughly independent of the particle size. The existence of the optimum airflow inlet velocity is due to formation of a thermal plume over the wound which acts as a protecting shield at low inlet velocities and is disturbed when the velocity goes over a certain limit. These features may be better understood by examining the near wound velocity and temperature distributions that are shown in Figure 3-a and 3-b for the inlet air velocities of 0.1 m/s and 0.2 m/s, respectively. As illustrated in the above mentioned figures, when the inlet velocity is sufficiently low, a thermal plume is formed on the wound due to the temperature difference between the wound tissue and the body skin. The thermal plume generates an upward flow over the wound which protects it from infection caused by deposition of airborne bacteria-carrying particles. As the airflow inlet velocity increases over the optimal value, in this case 0.1 m/s, the protecting thermal plume breaks down and particles deposition increases. As shown in Figure 4, breakage of the wound thermal plume is also clearly detectable from the dispersion of particles when the airflow inlet velocity increases from 0.1 m/s to 0.2 m/s. In addition, as can be seen in the figures, the direction of dominant flow near the wound changes from away from the wound to towards the wound as the inlet air velocity increases over the optimum limit. It is known that impaction is the main mechanism causing micro particles to deposit on the wound; thus, the amount of deposition depends not only on the concentration of particles over the wound, but also on the near wound velocity distribution. As the inlet air velocity
goes over the critical limit, airflow direction becomes predominantly towards the wound due to breakage of the protecting thermal plume and particles deposition increases considerably by the impaction effect. As a result, while the ventilation airflow in the operating rooms should be sufficiently high to remove particles from the wound area, it should not be too high to destroy the protecting wound thermal plume and to change the near wound velocity direction towards the wound. This conflicting needs of operating rooms ventilation system lead to existence of an optimal value for the inlet air velocity of the UCV systems. At too low inlet velocity, ventilation airflow is not sufficient to dilute the concentration of infectious airborne particles over the wound and particles deposition may be high. On the other hand, when the inlet airflow velocity is higher than the optimum value, the wound thermal plume breaks down and particles deposition increases due to the impaction effect, as was noted before. As a result, there is an optimum airflow inlet velocity that minimizes particles deposition on the wound. As the optimum velocity mainly depends on the near wound velocity distribution, it is almost independent of the size of released particles, as indicated in Figure 2.

To further clarify the role of the protecting thermal plume in preventing the wound from airborne particles, a special case is simulated for which the wound temperature is equal to the skin temperature so that the thermal plume is not formed. Figure 5 shows the velocity and temperature distributions near the wound in this case. It is seen that no plume is present and the airflow velocity is roughly towards the wound that causes the contaminant particles to deposit on the wound due to impaction. To quantify the effect of the thermal plume, in Figure 6, the deposition curves for 1 and 10 µm particles in the absence of the thermal plume are compared with the results for a case in which the thermal plume is formed. This figure indicates that without a temperature difference between the wound tissue and the body skin, increase of the inlet air velocity decreases the particles deposition monotonically due to the
dilution of the particle concentration over the wound. In this case, there is no optimum value for the airflow inlet velocity for minimum deposition. Figure 6, however, shows that the particles deposition decreases sharply with increase in the inlet velocity up to about 0.12 m/s and then remains roughly constant with further increase of the inlet velocity. This implies that the effect of the wound thermal plume is more pronounced at low inlet air velocities when it acts as a shield to protect the wound from large deposition of infectious airborne particles. As the ventilation airflow becomes quite high, the effect of the wound thermal plume diminishes and the results become roughly the same for both cases.

Surgical luminary is one of the most important upstream obstacles in an operating room which can considerably influence the airflow pattern and the resultant particle dispersion due to both generating an upward thermal plume and obstructing the ventilation airflow. In the modeling, the strength of the thermal plume formed over the lamps mainly depends on the thermal boundary condition used for the lights surfaces. In Figure 7, the effect of the surgical luminary boundary condition on the particle deposition over the wound is shown for two constant temperature boundary conditions, namely 350 K and 400 K, and two constant heat flux ones, namely 100 Watts and 200 Watts. As depicted in the figure, the thermal boundary condition has considerable influence on the particle deposition curve, both on the deposition percentage values and on the optimum inlet air velocity. The negative effect of the generated thermal plume is such strong that it may increase the optimum inlet velocity up to 0.14 m/s. The influence of the surgical lights thermal plume on the indoor airflow pattern may be better described by Figure 8 which shows the velocity and temperature distributions near the lamps under various thermal conditions when the inlet air velocity equals 0.10 m/s. As depicted in the figure, the thermal plume causes an upward flow over the lights which may disturb the upstream airflow pattern. To achieve the best ventilation performance, the inlet airflow should
have enough momentum to overcome the upward flow, corresponding to the inlet air velocity of 0.14 m/s for the case (c) which has the strongest thermal plume. Besides the thermal plume generated over the lights which may disturb the upstream airflow, a surgical lamp may also directly act as an obstacle and block the ventilation airflow. In this respect, the shape of a luminary and its location considerably influence the airflow pattern and the dispersion of infectious airborne particles. The effect of surgical lamps location and geometry on the wound infection risk can be better described referring to Figure 9 which shows the percentage of 10 μm particles deposited over the wound. As depicted in the figure, although the luminary shape and location has almost no major influence on the optimum inlet air velocity, improper location of a light may considerably increase the deposition of particles, even up to 100%. Independency of the optimum inlet velocity from the geometry and location of the lamps is mainly due to the fact that these parameters have negligible influence on the near wound velocity and temperature distributions and they are almost similar to Figure 3-a. This feature indicates that geometric characteristic of luminaries are not as important as their thermal characteristics in impressing the wound thermal plume and in changing optimum ventilation air velocity, accordingly. Unfavorable influences resulting from the surgical lamp location may be remarkably controlled using lights with a well-designed geometry. The important point is that the position of lights is almost out of a ventilation expert responsibility and it is usually adjusted by the surgeon to properly illumine the surgical wound. As a consequence, the design of ventilation system should be in a manner to minimize the negative effects resulting from adjustment of lamp location by the surgical staff. A new method to improve the air quality under UCV systems is using partitions over the surgical zone, as proposed by NHS [12]. However, the effect of the partitions on the airflow distribution and the particle dispersion through operating rooms has not been evaluated in details, to a best of authors’ knowledge. As shown in Figure 1, the partitions are 1 m in height
and are mounted around the ultra-clean area, either permanently or temporarily. The main effect of the partitions is to direct the ventilation airflow towards the surgical table and to prevent its momentum from decay due to expansion effect. As a consequence, particles concentration over the wound is diluted and their deposition decreases accordingly. However, this feature is not strong enough to impose any considerable effect on the wound thermal plume and the near wound temperature and velocity distributions are almost similar to Figure 3-a. Quantitative effect of using the partitions on the particles deposition may be extracted referring to the deposition curves in Figure 10. As presented in the figure, installation of the partitions decreases the deposition of airborne particles on the wound due to providing cleaner environment over the wound. However, the partitions have no considerable effect on the optimum inlet air velocity, as discussed previously by looking at the near wound velocity and temperature distribution.

4. Conclusions

In this study, the effect of the UCV system characteristics on the operating room ventilation system performance and on the risk of wound infection was investigated numerically. The results may be summarized as follows:

1. Due to a relatively large size of the studied particles, the contribution of Brownian motion in the particles deposition is almost negligible. Accordingly, the main deposition mechanisms are impaction and gravitational sedimentation, while the effect of the latter one vanishes rapidly as the particle size goes below 1 \( \mu \text{m} \).

2. Formation of a thermal plume over the wound, due to a temperature difference between the wound tissue and the body skin, causes an optimum value for the inlet air velocity which minimizes the particles deposition. Over the optimum inlet velocity, the protecting
thermal plume is destroyed and the deposition of particles increases because of impaction effect.

3. Specification of the surgical luminaries has considerable effect on the wound infection risk and on the optimum inlet air velocity both due to generating an upward thermal plume and obstructing the ventilation airflow. However, the thermal characteristics of the lamps is more critical than their geometric characteristics in impressing the airflow pattern and the deposition of airborne particles over the wound. The negative effects of the improper location of a surgical lamp can be minimized using luminaries with a well-designed geometry.

4. Mounting partitions in the UCV systems decreases the percentage of deposited particles over the wound and may reduce the risk of wound infection. However, its effect on the near wound velocity and temperature distribution is not strong enough to make a remarkable influence on the wound thermal plume and on the optimum inlet air velocity.

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Table Captions

Table 1. Coefficients and source terms of the flow governing equations, Equation (1)
Figure Captions

Figure 1. Geometric characteristics of the operating room model.

Note: Operating room, $6.3 \times m \times 6.3 \times 3 m$ [49]; UCV system inlet: $2.8 \times 2.8 m$ [12]; UCV system outlets, $0.6 m \times 0.3 m$; Surgical wound: $0.3 m \times 0.3 m$ [7]; For other dimensions and all distances refer to DIN 4799 [50].

Figure 2. The effect of airflow inlet velocity and particle size on the deposition of particles over the wound.

Figure 3. The effect of airflow inlet velocity on the near wound velocity and temperature distributions: (a) 0.10 m/s; (b) 0.20 m/s.

Figure 4. The effect of airflow inlet velocity on the airborne particles dispersion: (a) 0.1 m/s; (b) 0.2 m/s.

Figure 5. The effect of the wound thermal plume on the near wound velocity and temperature distributions.

Figure 6. The effect of the wound thermal plume on the particles deposition over the wound.

Figure 7. The effect of the surgical light boundary condition on the near lamp velocity and temperature distributions: (a) constant temperature of 350 K; (b) constant temperature of 400 K; (c) constant heat flux of 100 W; (d) constant heat flux of 200 W.

Figure 8. The effect of the surgical light boundary condition on the particles deposition over the wound.
**Figure 9.** The effect of geometric characteristics of the surgical lights on the particles deposition over the wound.

**Figure 10.** The effect of the partitions on the particles deposition over the wound.
Table 1.

<table>
<thead>
<tr>
<th>Equation</th>
<th>$\phi$</th>
<th>$\Gamma_\phi$</th>
<th>$S_\phi$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Continuity</td>
<td>1</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Momentum</td>
<td>$V$</td>
<td>$\mu_{\text{eff}}$</td>
<td>$-\nabla p + \rho g \beta (T - T_\infty)$</td>
</tr>
<tr>
<td>Energy</td>
<td>$c_p T$</td>
<td>$\mu_{\text{eff}} / Pr$</td>
<td>-</td>
</tr>
<tr>
<td>Turbulent kinetic energy</td>
<td>$k$</td>
<td>$\mu_{\text{eff}} / \sigma_k$</td>
<td>$P_{\text{t}} - \rho \varepsilon$</td>
</tr>
<tr>
<td>Turbulent kinetic energy dissipation rate</td>
<td>$\varepsilon$</td>
<td>$\mu_{\text{eff}} / \sigma_{\varepsilon}$</td>
<td>$\varepsilon (C_1 P_k - C_2 \varepsilon) / k$</td>
</tr>
</tbody>
</table>

**Note:** $\mu_{\text{eff}} = \mu + \mu_t$  
$\mu_t = \rho C_\mu k^2 / \varepsilon$  
$C_\mu = 0.0845$, $C_1 = 1.42$, $C_2 = 1.68$
Figures

Figure 1
Figure 3
Figure 5
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Figure 8
Figure 9
Figure 10
Biographies

**Behrang Sajadi** is an Assistant Professor in School of Mechanical Engineering at University of Tehran, Iran. His current research interests include aerosol mechanics, indoor air quality, energy modelling, heat transfer enhancement in refrigeration systems, and novel HVAC and refrigeration systems.

**Mohammad Hassan Saidi** is a Professor in School of Mechanical Engineering at Sharif University of Technology, Iran. His current research interests include MEMS, heat transfer enhancement in boiling and condensation, modeling of pulse tube refrigeration, vortex tube refrigerators, indoor air quality and clean room technology, energy efficiency in home appliances, and desiccant cooling systems.

**Goodarz Ahmadi** is a Professor of Mechanical and Aeronautical Engineering at Clarkson University, NY. He has been awarded the title of “Clarkson Distinguished Professor". Some of his research interests include multiphase and granular flows, three-phase slurry flows, aerosols, micro-contamination control, turbulence modeling, stability of fluid motions, continuum mechanics, nonlinear random vibrations, and earthquake engineering.