Theoretical and Experimental Investigation of the Thermal Effects Within Body Cavities During Transendoscopical CO₂ Laser-Based Surgery

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Background and Objectives: The recent development of flexible hollow waveguides for MID-IR lasers may be utilized transendoscopically to ablate selectively neoplastic, superficial tissues within body cavities. Study goals are to investigate theoretically and experimentally heat distribution and thermal response of cavity lining, during CO₂ laser minimally invasive surgery (MIS), and to thermally optimize the procedure under practical conditions. Study Design/Materials and Methods: Mathematical model was developed to predict temperature distribution along cavity lining. Experimental setup was built, including all the necessary components for a fully feedback-controlled MIS, i.e., laser generator, gas insufflating system, surgical suction, and infrared imaging feedback mechanism, all controlled by central PC-based program. Thermal images of cavity lining were recorded and analyzed throughout varying conditions. Results: Thermal gradients along the cavity lining, during and after the laser irradiation, were obtained mathematically and experimentally. Diverse modes of heat dispersions were observed, as well as the relative contributions of user-controlled parameters to the maximal heat of cavity lining. The software-controlled setup has demonstrated the capacity to instantly manage varying conditions, by which it automatically protects cavity lining from getting overheated. Conclusions: Analytical predictions and experimental measurements were highly correlated. The software-controlled system may serve a powerful tool to control thermal side effects during MIS within body cavities. Lasers Surg. Med. 35:18–27, 2004. © 2004 Wiley-Liss, Inc.

Key words: laser–tissue interaction; endoscopy; infrared waveguides and thermal imaging bundle

INTRODUCTION

Among the medical lasers in use, carbon dioxide (10.6 μm) and Er:YAG (2.94 μm) lasers are the most highly absorbed in water and therefore in soft tissues [1–3]. Hence, considered best for treating such cases as mucosal gastric cancer [4,5] and Barrett Esophagus [6–12], where ablating superficial tissue layer is needed. These two Mid-Infrared lasers are guided by optical hollow waveguides [13–19] (developed by our group as well as by others). At the present, neither flexible hollow fibers, nor solid core mid-IR fibers, are used in endoscopical applications. Consequently, the unique benefits of Er:YAG laser and the most commonly used CO₂ laser are not yet applied endoscopically to treat the lining of body cavities.

To make this feasible, further development is required, taking into consideration the environmental conditions. Irradiating tissue within a closed space, using wavelengths that are mostly absorbed in water will cause the dispersion of hot steam that may thermally damage the surrounding tissue. Previous studies have investigated the temperature rise during laser–tissue interaction along the irradiated region [20–26]. This study focuses on minimizing the thermal side effects during laser–tissue interaction within body cavities, i.e., on hotspots and the temperature distribution along cavity’s lining other than the irradiated tissue. Subsequent sections exhibit the exploration of these side effects, theoretically and experimentally, as well as the management of an MIS procedure [27] via a software-controlled system, developed in our lab.

MATERIALS AND METHODS

This study incorporates analytical and experimental investigations. The theoretical model addresses some of the thermal effects observed in the experiment. It provides an analytical tool for predicting the maximum reachable
surface temperature of healthy tissue in a cavity within which a laser surgery is conducted.

The Theoretical Model

Being primarily comprised of water, a human body tissue is a highly effective radiant heat absorber in the infrared spectrum. Its opaque characteristics are slightly higher than those of water owing to pigmentation. When irradiated by a CO2 laser beam, the entire beam radiant power is absorbed within an extremely thin tissue surface layer. For a CO2 laser of wavelength \( \lambda = 10.6 \ \mu \text{m} \) the corresponding tissue spectral absorbance coefficient is \( a = 600 \ \text{cm}^{-1} \).

Consequently, 67% of the incident CO2 laser radiant power is absorbed within a depth \( d = a = 0.017 \ \text{mm} \), which defined as the optical penetration depth. Within this layer, the radiant energy flux decreases exponentially over depth. The absorbed radiant heat is converted to sensible and latent heat. The former raises the tissue temperature up to the boiling point, the latter provides the vaporization latent heat. During this process, heat is conducted from the ablated surface towards the underlying colder tissue space. However, if the laser beam intensity is sufficiently high (well above the threshold level), surface tissue ablation dominates the process and prevents any significant heat conduction beyond the optical penetration depth.

In pulse laser surgery, to ensure that tissue destruction would be confined within a thin surface layer, it is advisable to operate with a pulse duration time that is shorter than the conductive thermal relaxation time. It essentially limits the tissue destruction to depths that are on the order of the optical penetration depth. The thermal relaxation time is determined simply by equating the thermal conductive diffusion depth \( l = \sqrt{\frac{\pi a t}{D}} \) to the optical penetration depth \( d \); where \( a_t \) is the tissue thermal diffusivity and \( t \) is the time. Laser pulse duration shorter than the relaxation time \( t = \frac{l^2}{a_t} \) ensures that heat diffusion by conduction would not have time to reach and damage tissue beyond the optical penetration depth.

In continuous laser surgery, tissue destruction can be controlled by the laser intensity. Owing to, both the poor tissue thermal conductivity and strong infrared absorption properties, it is possible to limit the tissue destruction depth. With sufficient laser power, the rate at which the ablated surface recedes could match the thermal diffusion rate into the underlying tissue. This feature can be used to limit unnecessary collateral healthy tissue damage. When the rate at which the crater floor recedes matches the thermal diffusion rate, on moving coordinates, steady state conditions are established underneath the ablated crater floor. It means that for an observer standing on the moving floor, the temperature distribution underneath the crater floor remains constant. If the floor velocity is \( v \), it would recede a distance \( \delta_0 \) in a time interval \( t_0 = \frac{\delta_0}{v} \). For such steady state conditions, the conductive heat diffusion would have to penetrate a distance \( \delta \) in the same time interval, which is on the order of

\[
\delta_0 \approx \sqrt{2a t_0} = \sqrt{2a} \frac{\delta_0}{v}
\]

The thermal penetration depth seen underneath the ablated crater floor is therefore approximately

\[
\delta_0 = \frac{2t}{v}
\]

Under those steady state conditions, the movement rate of the crater floor (air/tissue interface) is obtained from the ratio between the laser power \( W \) and the tissue specific heat necessary for its vaporization from a normal body temperature

\[
v = W [\rho_l A (c\Delta T + h_{fg})]^{-1}
\]

where \( A \) denotes the laser beam cross section area, \( \Delta T \) the boiling to normal tissue temperature difference, \( \rho_t \) the density, \( c \) its specific heat, and \( h_{fg} \) its vaporization latent heat at the boiling point. The laser power that is needed to limit the thermal penetration depth so that it remains equal to the optical penetration depth is therefore

\[
W = a A \Delta T \rho_t (c\Delta T + h_{fg})
\]

For a laser beam of 1 mm diameter, the power required to limit the thermal penetration depth to \( \delta_0 = 0.017 \ \text{mm} \) is approximately 18 W. Half that power would entail a thermal penetration depth that is twice as large. Nonetheless, it would be still as thin as 30 \( \mu \text{m} \). In this respect, it is worth noting that most CO2 laser surgeries are conducted with power levels that range from 2 to 15 W. For simplicity, in that calculation and all subsequent calculations, the tissue thermal properties are assumed identical to those of water.

A well-controlled CO2 laser operation minimizes any unnecessary heat conductance and storage in the ablated crater surrounding tissue. Most of the radiant laser power is aimed and consumed by the evaporation process. That power is subsequently released out of the crater in the form of an emerging vapor plume. It consists of both the threshold sensible heat and the vaporization latent heat. Radiation leaving the crater by means of reflection, scattering, or emission is insignificant. This is a consequence of, both the tissue strong absorption characteristics and relatively low temperature (\( T < 100 \)\(^\circ\)C).

As previously indicated, during laser surgery within a cavity, such as the stomach, the vapor plume escaping the crater can potentially damage the cavity lining. To analyze the possibility of such damage, it is necessary to examine the thermal interaction between the vapor and the cavity surface. In this respect, it is important to address the geometry of both the vapor plume and the cavity.

Laboratory experiments indicated that in the absence of forced vapor suction, the vapor plume rises, as it emerges from the ablated crater, owing to its positive buoyancy. The latter is caused by, both the lower species density (H\(_2\)O) and higher temperature of the vapor plume as compared to the cavity ambient air. The experiments were conducted underneath a hemispherical dome. At the dome base center, tissue was ablated by a CO2 laser (see Fig. 1). It was observed, as described in “Experimental stages,” that the
vapor influx, combined with, both the plume buoyancy and the dome geometry induced a torus-like circulatory pattern underneath the dome. While rising in the center, the plume gradually mixed with the adjacent circulating ambient air. In effect, this mixing is the cause for the observed continuous expansion of the ascending plume cross section. Once reaching the dome peak, the plume spread radially and was induced to flow downwards in parallel to the dome walls, towards the dome base. During this movement, the vapor plume heats the dome wall by a process of natural convection heat transfer. During testing, attention was directed towards the monitoring of dome hottest spot temperature time history. This is the equivalent of monitoring the hot spot temperature rise of healthy tissue during laser surgery. The purpose is to avoid dangerously long healthy tissue exposure to elevated temperatures. In a safe laser surgery procedure, once the maximum allowable temperature/time limit is reached at the hot spot, the laser surgery must be halted, and if necessary resumed after an adequate cooling period.

In the test, the dome hottest spot is at the peak, where the rising plume first comes in contact, prior to its cooling as it moves along the dome wall. Numerous assumptions are introduced to facilitate the evaluation of the hot spot temperature rise. A lengthy and rigorous analysis of the problem is unwarranted, since there could be a wide range of actual cavity geometries. The dome geometry is essentially a representation of any cavity by a single characteristic dimension. Through the evaluation of the maximum to average heat flux ratio along the dome wall, it would be possible to estimate the hot spot temperature of any cavity subject to its characteristic dimension. Along that approach, only characteristic temperatures and velocities are considered, to account for the influence of various energy and momentum transport mechanisms.

A plume rising from a crater of radius \( b_o \) expands, by mixing with the ambient air \([28]\), to a radius \( b \). After traveling vertically across the dome radius \( R \), the plume radius is

\[
b = b_o + 0.144R \tag{5}
\]

As aforementioned, at the dome summit the plume is forced to change course and expands radially. Assuming that the local momentum loss through this flow diversion is negligible, the boundary layer thickness \( \delta \) into which the diverted radial flow moves is obtained from the mass flow continuity equation, and is

\[
\delta = \frac{xb^2}{2n} = \frac{b}{2} \tag{6}
\]

The plume characteristic average velocity at the dome peak is

\[
V = 0.7466 \left( \frac{55.66Wq_f\beta}{\rho_c\beta y} \right)^{1/3} \tag{7}
\]

where \( q \) is the laser power, \( g \) the gravitational acceleration, \( \beta \) the gas temperature expansion coefficient, \( c_p \) the gas constant pressure specific heat, and \( y \) the vertical distance from the virtual point source \((y = b/0.144)\).

The hot plume leaving the ablated crater enters the revolving vapor cloud at the base center. Once reaching the dome summit it began to slide downwards along the dome curved wall. The boundary layer within which this flow travels is assumed to be of constant thickness. This assumption is drawn from the fact that a free convection boundary layer thickness is usually weakly dependent on the longitudinal coordinate. For the dome flow, diffusive viscous effects would tend to slow the flow and thereby expand the boundary layer thickness. Opposed to that is the boundary layer tendency to contract owing to geometrical flow expansion. These effects, at least partially could offset each other. Furthermore, the overall circular flow pattern is expected to adjust to the dome shape and thereby keep the boundary layer streamlines parallel to the wall. With the assumption of a uniform boundary layer thickness, the vapor characteristic longitude velocity \( u \) near the wall would depend only on the polar (latitude) angle \( \theta \); as measured from the dome summit, where \( \theta = 0^\circ \), to the base, where \( \theta = 90^\circ \). That dependence could be found from the boundary layer continuity equation

\[
d(u \times \sin \theta) = 0 \tag{8}
\]

If \( \theta_o \) is the angle where the plume enters the boundary layer near the dome summit, then the corresponding boundary condition would be

\[
u(\theta_o) = V \text{ at } \theta_o = \tan^{-1}(0.144) \tag{9}
\]

Integration of the equation, subject to the boundary condition yields

\[
u = \frac{u_o \sin(\theta_o)}{\sin(\theta)} \tag{10}
\]

Similarly, a heat transfer balance on an element volume \( 2\pi R^2\sin(\theta)\delta \theta \) could reveal the dependence of the
boundary layer characteristic temperature $T$ on polar angle $	heta$, therefore

$$\frac{k}{\delta} (T - T_W) 2\pi R^2 \sin(\theta) d\theta = -d(\rho c_p u T \delta 2\pi R \sin \theta)$$ (11)

where $k$ is the gas thermal conductivity, and $T_W$ the dome surface temperature. For simplicity, the surface temperature is considered constant in the study of the heat flux distribution. The temperature builds up with time will be considered specifically at a later stage of the analyses. The permissible safe temperature rise is small enough not to affect much the heat flux distribution. Equation (11) satisfies the requirements that the heat transferred to the dome surface is drawn from the convect heat within the boundary layer. Substitution of the velocity term into Equation (11) and differentiation gives

$$\frac{dT}{T - T_W} = \frac{2R}{\delta^2 u_0 \sin \theta_0} d(\cos \theta)$$ (12)

where $\alpha$ is the gas thermal diffusivity. The corresponding boundary condition is

$$T(\theta_0) = T_0$$ at $\theta = \theta_0$ (13)

$T_0$ is the plume characteristic temperature at the dome summit. In that representation, it is assumed that at the dome summit, the plume entire energy is in the form of sensible heat. This assumption is based on the expected vapor condensation that would occur owing to the mixing with the cooler ambient air during the plume ascent. Integration of Equation (12) subject to its boundary condition yields

$$\frac{T - T_W}{T_0 - T_W} = \exp \left[ \frac{2R}{\delta^2 u_0 \sin \theta_0} (\cos \theta - \cos \theta_0) \right]$$ (14)

The maximum heat flux $q_{\text{max}}$ is at the dome summit (where $\theta = \theta_0$) and is equal to

$$q_{\text{max}} = \frac{k}{\delta} [T_0 - T_W]$$ (15)

The average heat flux is by definition

$$\bar{q} = \frac{\int q \, dA}{\int dA} = \frac{\int q \, dA}{\int 2\pi R^2 \sin \theta d\theta}$$ (16)

Substitution of the temperature and integration yields

$$q = \frac{k}{\delta} [T_0 - T_W] \sin \theta_0 \exp(-\kappa \cot \theta_0) \left[ \exp \left( \frac{k}{\sin \theta_0} \right) - 1 \right]$$ (17)

where the parameter $\kappa$ is defined as

$$\kappa \equiv \frac{2R}{\delta^2 u_0} = \frac{4\pi R}{(b_0 + 0.144R)^2 V}$$ (18)

The ratio $\chi$ of the maximum to average heat flux is therefore

$$\chi = \frac{q_{\text{max}}}{\bar{q}} = \frac{\kappa}{\sin \theta_0} \exp(\kappa \cot \theta_0) \left[ \exp \left( \frac{k}{\sin \theta_0} \right) - 1 \right]^{-1}$$ (19)

Since $\theta_0$ is a small angle, the fluxes ratio expression can be well approximated by

$$\chi = \frac{k}{0.14} \left[ 1 - \exp \left( -\frac{k}{0.14} \right) \right]^{-1}$$ (20)

The heat fluxes ratio of a $q = 12$ W laser ablating a crater of radius $b_0 = 0.8$ mm under a dome of radius $R = 40$ mm, according to Equation (19), is $\chi = 1.88$. Experiments conducted in our laboratory revealed a ratio of $\chi = 2$. Therefore, introduction of an empirical correction factor that corresponds to $\phi = 1.065\chi$ seems to be appropriate. Before its incorporation, sensitivity analyses were conducted on the dependence of Equation (20) on its constituents. For instance, an increase of 20% in either, the cavity radius or the laser power entails corresponding reductions of $\chi$ by some 6 and 3%, respectively. This weak dependence of $\chi$ on its main variables justifies the incorporation of that factor. Consequently, the maximum heat flux could be calculated according to

$$q_{\text{max}} = \phi q = \frac{3.8kq}{2\pi R^2} \left[ 1 - \exp \left( -\frac{k}{0.14} \right) \right]^{-1}$$ (21)

This heat flux would be the driver of the healthy tissue temperature rise. As previously discussed, the poor tissue thermal diffusivity limits the penetration depth of the temperature rise. The heated tissue layer underneath the cavity lining would be very shallow. Heat conducted into such a thin layer will always be in a direction perpendicular to the surface. The curvature of the surface does not affect the temperature response of the cavity lining to the heat flux. To demonstrate that, consider a heated area of spherical cavity. The temperature response in the wall of the cavity is obtained from the solution of the conduction equation in polar coordinates [29], and is

$$T - T_1 = \frac{Rq}{k_t} \left[ \text{erfc} \left( \frac{r - R}{2\sqrt{\kappa t}} \right) - \exp \left( \frac{r - R}{R} \pm \frac{2\pi R}{\sqrt{\kappa t}} \right) \right]$$ (22)

where $q$ is the local heat flux, $k_t$ the tissue thermal conductivity, $T_1$ the initial temperature, and $r$ the radial coordinate. The tissue surface temperature is obtained simply by substituting $r = R$, which yields

$$T_W - T_1 = \frac{Rq}{k_t} \left[ 1 - \exp \left( \frac{2\pi R}{\sqrt{\kappa t}} \right) \text{erfc} \left( \frac{\sqrt{\kappa t}}{R} \right) \right]$$ (23)

This expression can be further reduced to reflect the cavity surface temperature response to surgery durations of few seconds. For a cavity of few centimeters radius surrounded by walls with a poor thermal diffusivity, such as a tissue, the following inequality applies

$$\frac{\sqrt{\kappa t}}{R} \ll 1$$ (24)

It is essentially an indication that the temperature penetration depth after a few seconds is insignificant as
To avoid undesirable healthy tissue damage (i.e., inflammation or cell necrosis), it is necessary to limit the maximal temperature rise to 46°C [1,30]. Equation (26) can be directly applied to calculate the corresponding maximum time, \( t_{\text{max}} \), within which the laser collateral heating is tolerable. That time can be calculated from the substitution of that equation into Equation (25) for spherical domes. Substitution of that equation into Equation (26) provides the maximal cavity lining temperature

\[
T_{\text{max}} - T_1 = \frac{q_0 q}{\pi^{1.5} R^2 k_1} \sqrt{\frac{a_1 t}{R}}
\]  

(26)

For the heating phase, when \( t \ll t_{\text{max}} \), Equation (26) applies.
IR camera was placed beneath cavity cradle, focused on the top hemisphere through a nylon-wrapped window at the bottom hemisphere and set to 0.95 emissivity [38,39]. Laser beam diameter was measured as 1.6 mm, based on the “razor-blade” technique [40]. The average power density (APD) was then calculated as the power/($\pi r^2$) or power $\times$50, i.e., 1 W power resulted in 50 W/cm$^2$ APD. The hollow optical waveguide was attached to the laser arm via a custom-made optical coupler and a carved adapter. The laser beam was pointed at the tissue samples at the bottom. Fiber, suction and insufflation outlets where threaded through corresponded holes in cavity's roof (Fig. 3).

**Experimental Stages**

Our first stage of research was to explore the diverse modes of the vapor plume distribution, under varying conditions, from which we could learn the hotspots location, along cavity lining. For that purpose, transparent (nylon) domed roofs were constructed. Smoke distribution was visually recorded (Fig. 4).

Knowing where hottest areas are, we used plastic domed roofs, with tissue samples placed at the center (Fig. 3), to record the thermal effects via IR camera, under varying parameters, e.g., laser power (0–15 W), pulse duration (0–10 seconds, CW), insufflating gas flow rate (0–15 lpm), suction volume (0–50 mmHg), airflow through the hollow waveguide (0–3 lpm).

Finally, the software-controlled setup was tested. It monitored the maximal temperature of the surroundings, analyzed constantly from the IR images, and regulated automatically the parameters of laser, suction or insufflation, to maintain the temperature within the selected range. Regulation limits and increments (rate) were tunable. It further enabled us to document the values of temperature and parameters versus time.

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**Fig. 3.** Cavity-like phantom; internal view through IR camera window; suction, insufflation and fiber outlets adjacently located, same as in endoscope distal tip. [Figure can be viewed in color online via www.interscience.wiley.com.]

**Fig. 4.** Vapor pattern underneath the transparent dome: Plume (left); circulatory (right). [Figure can be viewed in color online via www.interscience.wiley.com.]
RESULTS

Analytical Versus Measured Results

The maximal temperature response along cavity lining predicted by Equations (26) and (27) was tested experimentally. Figure 5 demonstrates a comparison between predicted and measured data. In both, the continuous laser irradiation of 600 W/cm² is stopped after 12 seconds. During the irradiation period the maximal temperature of the surrounding tissue had climbed to approximately 37 °C, by which a thermal gradient of 13 °C was built-up, and then declined exponentially to the approximate rate of 29 °C within 13 more seconds.

Heat Distribution Modes

In the absence of gas flow and vacuum, two types of dispersion patterns were observed: plume and jet. The plume column headed straight up whereas the jet was directed contrary to the laser beam (i.e., toward the fiber). The jet turned to be dominant beyond 750 W/cm² power densities. In both cases, right after vapor column hit cavity lining it formed a mushroom-like distribution, and then got turbulent (Fig. 4). Consequently, hotspots along cavity lining were top centered in the plume case, around fiber entrance hole in the jet case or in-between, depending on laser power and cavity radius. Inside these hotspots, mucosal folds (Rugae) were the ones heated most. It was also noticed that the jet mode caused the penetration of vapor into fiber’s hollow core, thus heated the fiber and the tissue surrounded it and rapidly attenuated its power transmission. Therefore, the fiber was inserted diagonally into the cavity.

Insufflating gas, injecting airflow through the hollow fiber, and activating the suction caused more homogeneous heat dispersion or none, depending on the relative volume of each. Neither plume nor jet was observed at any of these cases. Gas insufflating helped to “blur” hotspots (i.e., reduced the maximal temperatures) not only by circulating the vapor but also by straightening tissue folds.

Maximal Heat Build-Up

The graphs in Figures 6 and 7 represent the maximal temperature of cavity lining under varying conditions and contributions of each of the parameters. Figure 6 demonstrates the thermal gradients obtained at 15 W (750 W/cm²), continuous induction, for 8 seconds. The least gradient of 0.5 °C resulted at 3 lpm airflow, injected through the hollow waveguide and 50 mmHg vacuum power after 8 seconds, a 1.5 and 4.7 °C gradients resulted in the sole presence of vacuum or airflow, respectively and the highest gradient of 8 °C was built-up in the absence of both (gasless conditions). Figure 7 exhibits the fairly linear relation between thermal gradient and laser power. Figure 8 puts on display samples of the thermal images, recorded from the surrounding tissue throughout altering conditions. These temperature maps demonstrate graphically how
airflow and vacuum can be manipulated to blur or to prevent the formation of hotspots along cavity lining.

**Testing the Software-Controlled System**

Figure 9 shows how the computerized system monitors the maximal temperature and manipulates the parameters when temperature exceeds a selected value, to avoid irreversible thermal side effects. In this example, the CO$_2$ flow was regulated. The maximal flow was set to 15 lpm and the temperature upper limit to 46°C. A continuous laser irradiation of 2 W (100 W/cm$^2$) has elevated the maximal temperature of the surrounding tissue. When the temperature crossed 46°C, the gas flow was automatically activated and increased (at a user-defined rate) via controller and proportional valve, till the maximal temperature declined back below the limit. The flow rate was then stabilized. In general, higher limit of maximal gas flow enabled handling higher temperatures, whereas greater flow increment obtained faster regulation, however tuning of the final flow rate and the temperature turned rougher. Regulating automatically suction volume, laser power or transmission duty cycle had exhibited principally similar capabilities to control the rate of temperature stabilization and the precision of the steady state temperature.

**DISCUSSION**

Analytical model and experimental measurement were highly correlated. The agreement between the predicted
had declined back and stabilized on 46°C; after first two flow automatic activations, flow was manually stopped to enable heat development. Then flow was reactivated and reached its top limit of 15 Ipm. Temperature feedback via IR imaging fiber bundle.

**FUTURE INVESTIGATION**

Stepping into animal trials, we intend to utilize the control system, by applying the laser energy transendoscopically, via flexible fibers, and obtaining the thermal feedback via IR imaging fiber bundle.

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