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Comparison of Pulsed CO_2 Laser Ablation at 10.6 μm and 9.5 μm

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Background and Objective: **The pulsed CO_2 laser has received attention because of its successful application to dermatologic surgery and burn debridement surgery. Despite impressive results, tissue removal using pulsed CO_2 laser irradiation has not been optimized. We examined the ablation processes by performing mass removal and thermal injury experiments at wavelengths where tissue water is the primary absorber (10.6 μm), and where water and collagen have comparable absorption (9.5 μm).**
Study Design/Materials and Methods: **Samples of porcine reticular dermis were irradiated with 180-ns laser pulses at either wavelength. Tissue removal was measured using a digital balance. Thermal injury was assessed using a microscope with a calibrated reticle after hematoxylin and eosin staining.**
Results: **Tissue removal using 10.6-μm radiation resulted in a heat of ablation of 3,740 J/g, an ablation threshold of 1.15 J/cm^2, and a zone of thermal injury of 53 μm. By contrast, tissue removal using 9.5-μm radiation resulted in a heat of ablation of 3,330 J/g, an ablation threshold of 1.47 J/cm^2, and a zone of thermal injury of 34 μm. The differences in ablation threshold and thermal injury were statistically significant.**
Conclusion: **Pulsed CO_2 laser irradiation at 9.5 μm removes tissue more efficiently and with a smaller zone of thermal injury than at 10.6 μm. Lasers Surg. Med. 23:1–6, 1998.**

Key words: heat of ablation; laser ablation efficiency; explosive vaporization; thermal injury

INTRODUCTION

The carbon dioxide (CO_2) laser has been widely used in medical applications because of its ability to coagulate, incise, and excise biological tissue. Recently, the pulsed CO_2 laser has received attention because of its successful application to skin surgery and burn debridement. Although continuous wave (cw) CO_2 lasers were used in the 1980s and early 1990s, the high degree of thermal injury produced during ablation (200–1,000 μm) caused unpredictable results with an increased risk of scarring. High-energy pulsed CO_2 lasers or rapidly scanned cw lasers limit thermal diffusion during ablation and therefore decrease thermal injury to approximately 70–160

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μm [1]. Despite widespread clinical use, tissue removal with pulsed CO_2 laser irradiation continues to generate thermal injury in excess of that which is necessary to produce hemostasis, often causing temporary and sometimes permanent side effects such as hyperpigmentation, hypopigmentation, edema, and erythema (eg. [2,3]). While pre- and post-treatment therapies can reduce healing time [4], optimization of the laser system to control depth of thermal injury for surgical procedures would be beneficial.

Unlike infrared lasers, ultraviolet lasers tend to remove tissue cleanly, efficiently, and with minimal thermal injury [5,6]. Recent work has suggested that tissue chromophore may be responsible for the disparate effects produced by different laser wavelengths [7–9]. It has been observed that when collagen is directly targeted by choosing an appropriate wavelength such as 193 nm and 248 nm, material removal is consistent with a process of rapid surface vaporization [7]. However, when tissue water is targeted by using infrared wavelengths such as 2.79 μm and 10.6 μm, a delayed and more explosive ablation process is observed [8]. Because ultraviolet lasers produce high-energy photons, some investigators have postulated optical and photochemical phenomena to explain ablation processes occurring during ultraviolet laser ablation [10–15]. However, Edwards and co-workers have provided evidence that low photon energies can also produce clean and efficient ablation events by targeting the amide II band of collagen at 6.45 μm [9]. Although 6.45 μm lasers are not broadly available, other laser lines could be used to target collagen.

The CO_2 laser is capable of producing high-energy pulses at wavelengths in the 9–11 μm region. The absorption spectra of dry collagen [16] and water [17] within the CO_2 laser wavelength range are shown in Figure 1. At the laser wavelength used in most CO_2 laser applications (10.6 μm), the dominant chromophore is water, whereas at 9.5 μm, the absorption of collagen and tissue water are comparable. We therefore hypothesized that irradiation at 9.5 μm would affect a more efficient ablation process with less thermal injury. The purpose of this study was to test this hypothesis by comparing clinically relevant parameters such as ablation efficiency, ablation threshold, and depth of thermal injury for pulsed CO_2 laser irradiation at wavelengths 10.6 μm and 9.5 μm.

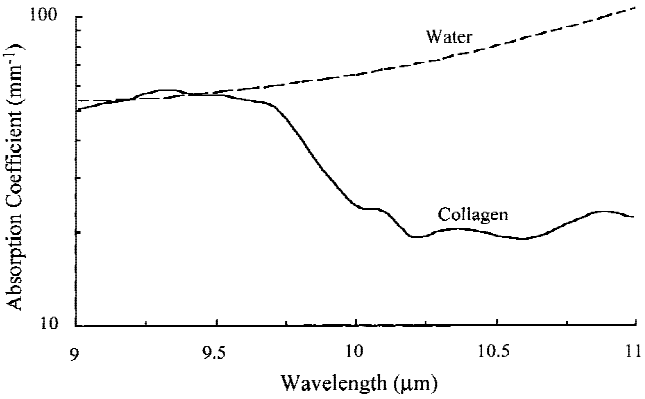

Fig. 1. Absorption spectra for dry collagen and water in the far IR region.

MATERIALS AND METHODS

Laser

A nitrogen-starved tunable transversely excited atmospheric (TEA) CO_2 laser (840, Lumonics, Kanata, Ontario) was used for all experiments. The laser wavelength was selected by tuning a diffraction grating at the rear of the laser cavity. The diffraction grating was calibrated using a CO_2 spectrum analyzer (Laser Craft, Santa Rosa, CA). The temporal width of the laser pulse consisted of a single 180-ns pulse (full-width half-maximum) with no observable tail as measured by a pyroelectric detector with a response time of 500 ps (P5-02, Molectron Detector, Portland, OR). The laser output was horizontally polarized with a pulse to pulse energy variation of less than 4%. A uniform portion of the laser beam was selected with a 15-mm aperture and focused onto the target by a 203-mm focal length ZnSe lens. Attenuators were used to vary the incident energy that reached the tissue target, and a helium-neon laser coupled with a beamsplitter was used to identify the ablation plane [18]. The spatial profile of the laser beam was measured by stepping through the ablation plane with a 300-μm-diameter aperture and measuring the transmitted energy using a pyroelectric detector (J3-09, Molectron Detector, Portland, OR). The $1/e^2$ diameters were calculated using a gaussian least-mean-squares fit and were 2.68 mm and 2.44 mm for wavelengths 10.6 μm and 9.5 μm, respectively. The difference in spot sizes were due to slight differences in laser mode and the refractive index of ZnSe at 10.6 μm and 9.5 μm wavelengths.

Tissue Removal Experiments

Sections of porcine reticular dermis approximately 4-mm thick were excised immediately postmortem and irradiated within 12 hours. Prior to irradiation, 6-mm biopsy punch samples were obtained and placed in a plastic holder on the pan of a digital balance (AE163, Mettler Instrument Corp., Hightstown, NJ) so that all ablated material would land off the pan [18]. Twelve pulses at 1 Hz were delivered in sequence to each tissue sample. The number of pulses was held constant at 12 to ensure sufficient mass removal but minimize tissue dehydration during the experiment. Mass loss data were acquired using a personal computer at 2.4 Hz using data acquisition software (LabView, National Instruments, Austin, TX).

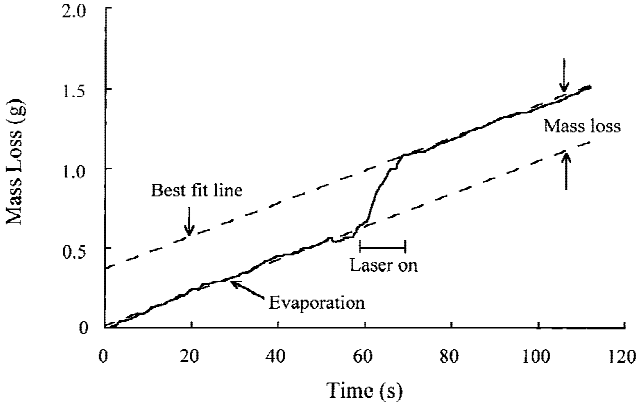

Fig. 2. Typical mass loss versus time curve for TEA CO_2 irradiation. Twelve pulses at 1 Hz were delivered to the porcine dermis target. Mass removal before and after laser irradiation is due to evaporation.

Thermal Injury Experiments

The experimental setup was identical with the tissue removal setup, except that a digital balance and computer were not used. The energy delivered to the tissue target was chosen based on two criteria. First, the pulse energy was normalized with respect to ablation threshold as determined from the tissue removal experiments. Second, the pulse energy was as high as possible without causing plasma formation. The radiant exposures that satisfied these conditions were 3.5 J/cm^2 and 4.5 J/cm^2 for wavelengths 10.6 μm and 9.5 μm, respectively. Fifteen tissue samples were irradiated at each wavelength, and the number of pulses delivered to each sample was held constant at 12. After laser irradiation, the tissue samples were fixed in formalin, set in paraffin, cut into 15-μm-thick sections, and stained with hematoxylin and eosin. Thermal injury was assessed by a pathologist who was blinded to the laser parameters used. Thermal injury was measured at the base of each crater using a calibrated reticle to eliminate edge effects.

RESULTS

Tissue Removal Experiments

Figure 2 shows a typical plot of the mass lost as recorded by the digital balance versus time. Before the tissue is irradiated, mass removal due to water evaporation from the tissue surface is seen. Once irradiation commences, an increased rate of mass removal is observed due to ablation. After irradiation is complete, mass removal continues due to evaporation. A linear regression analysis was performed on the mass loss data before and after ablation. The y-intercepts were calculated using the average of the slopes to yield a total corrected mass loss which is shown versus radiant exposure in Figure 3. Optical breakdown or plasma formation (visually observed as a bright white flash) was seen at radiant exposures above 3.7 J/cm^2 and 4.8 J/cm^2 for wavelengths 10.6 μm and 9.5 μm, respectively. A linear regression analysis was performed on the data for radiant exposures where optical breakdown was not observed. The resulting least-squares fit is shown in Figure 4. The slope of the line represents mass removed per unit energy delivered, and we define this as ablation efficiency (μg/J). The heat of ablation is the reciprocal of the ablation efficiency and is the energy required to remove a unit mass of tissue. The x-intercept is the minimum energy required to achieve material removal and is known as the ablation threshold (J/cm^2). A summary of the regression and ablation parameters is shown in Table 1. Although the difference in ablation efficiency at the two wavelengths was not quite statistically significant, repeated experiments exhibited the same bias. The difference in ablation threshold was statistically significant ($P = .005$).

Thermal Injury Experiments

Typical histological sections are shown in Figure 5. The difference between normal and thermally altered tissue is evident by the change in tissue staining. At 9.5 μm, the ablation results in a smoother crater with a more superficial zone of thermal injury when compared to 10.6 μm. The

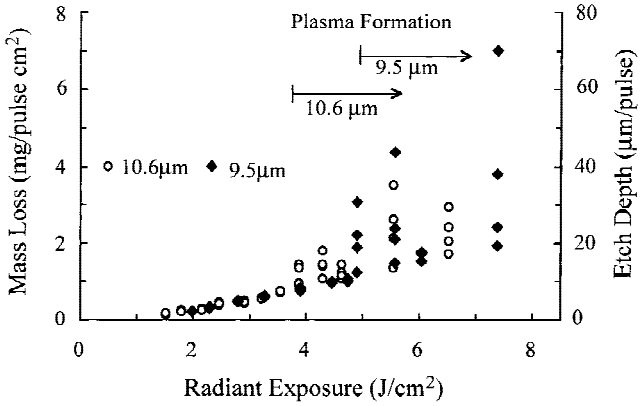

Fig. 3. Mass of porcine dermis removed per pulse per unit area versus laser radiant exposure by a TEA CO_2 laser at wavelengths 10.6 μm and 9.5 μm. The spread of the data at higher radiant exposures is due to plasma formation.

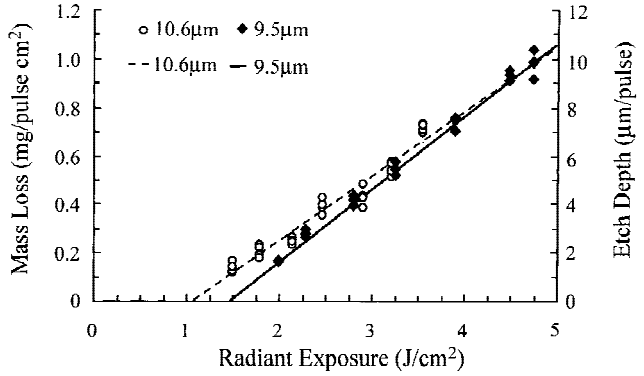

Fig. 4. Mass of porcine dermis removed per pulse per unit area versus laser radiant exposure by a TEA CO_2 laser at wavelengths 10.6 μm and 9.5 μm. Data are fit to a linear regression before optical breakdown was observed.

TABLE 1. Linear Regression and Ablation Parameters

Wavelength	10.6 μm	9.5 μm
No. of samples	28	28
Onset of plasma formation	3.7 J/cm²	4.8 J/cm²
Radiant exposure range	1.5–3.6 J/cm²	2.0–4.7 J/cm²
Correlation coefficient (r)	0.975	0.996
Slope	267 μg/J (242–292)[a]	300 μg/J (290–311)[a]
X-axis intercept	1.15 J/cm² (1.03–1.27)[a]	1.47 J/cm² (1.41–1.54)[a]

[a]95% Confidence interval.

mean thickness of thermal injury ± 1 standard deviation was 34 ± 9 μm and 53 ± 17 μm for TEA CO_2 irradiation at wavelengths 9.5 μm and 10.6 μm, respectively. The difference between these values was statistically significant (t-test, P = .002).

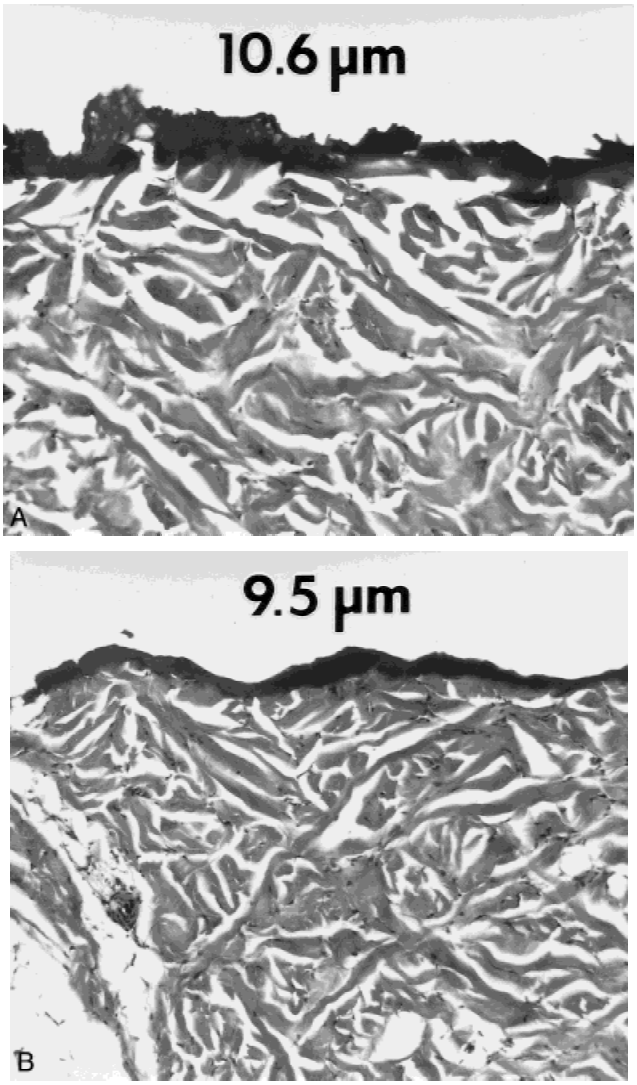

Fig. 5. Typical histology slides after 12 pulses of TEA CO_2 irradiation. The zone of thermal injury is approximately 60 μm and 30 μm for wavelengths 10.6 μm **(A)** and 9.5 μm **(B),** respectively. ×100.

DISCUSSION

This report describes tissue removal and thermal injury experiments performed on porcine dermis using TEA CO_2 laser irradiation at wavelengths 10.6 μm and 9.5 μm. At both wavelengths ablation appeared to be explosive, causing tissue fragments to collect on or near the ZnSe focusing lens. We had initially thought that by targeting collagen and tissue water equally using 9.5-μm radiation, a surface vaporization event would occur, whereas an explosive event would occur when only tissue water was targeted using 10.6-μm radiation. However, under our experimental conditions, it appears that sufficient laser energy was

available in the tissue water to produce explosive material removal at both wavelengths [8]. It follows that if collagen is more selectively targeted by using another wavelength, the energy absorbed by the tissue water will be small and may not be sufficient to cause explosive vaporization. Therefore, tissue chromophore, or more precisely, the ratio of absorption coefficients of the chromophores within the tissue volume, may govern ablation mechanism. This is presumably why ultraviolet irradiation is consistent with a process of rapid surface vaporization and not explosive vaporization [7]. However, the specific role of ablation mechanism on clinically relevant parameters such as ablation efficiency, threshold, and depth of thermal injury remains unclear.

The pulse duration used in this study was chosen to limit thermal diffusion out of the chromophore during irradiation, thereby allowing direct targeting or photothermolysis of the chromophore [19]. The time scale of thermal diffusion (τ_d) is given by $\tau_d \approx \delta^2/\alpha$, where δ is the characteristic size of the chromophore and α is the thermal diffusivity. Assuming the characteristic size of the chromophore (collagen fiber) is ≈ 1 μm, and the thermal diffusivity is comparable to water, the time scale for thermal diffusion is ≈ 7 μs. Therefore, to limit thermal diffusion during irradiation, the pulse duration must be less than 7 μs. One should note that the pulse duration used in these experiments is considerably different from pulse durations used in many current dermatologic procedures.

The question still remains as to what determines the efficiency of an explosive ablation process. Explosive vaporation can be produced by a rapid phase transition to vapor from mechanically unstable superheated tissue water at the spinodal limit [8,20–23]. When phase separation occurs, very high pressures are generated which are comparable to the ultimate tensile strength (UTS) of porcine dermis (8–10 MPa) [24]. It follows that tissues with low tensile strength will be less likely to impede material removal, thereby allowing a more efficient ablation event. In other words, when phase separation occurs, the efficiency of material removal will directly depend on the tissue UTS. During 10.6-μm irradiation, we speculate that the tissue structural matrix remains intact, and therefore produces a less efficient ablation process. However, during 9.5-μm irradiation, the tissue structural matrix is compromised or weakened by directly targeting collagen, and consequently produces a more efficient ablation process. These results are consistent with those of Walsh and co-workers who observed increased ablation efficiencies when irradiating tissues with lower ultimate tensile strength but equal optical absorption [25].

In our studies residual thermal injury appears to be inversely related to ablation efficiency. When compared to 10.6 μm, 9.5-μm irraidation resulted in a more efficient removal process with smaller residual thermal injury. This inverse relationship between ablation efficiency and depth of thermal injury is most easily explained by considering how the delivered laser energy is used. Laser energy must either contribute to the vaporization and ejection of ablated material or the heating of the remaining tissue, thereby producing thermal injury by denaturing tissue components. It follows that the more efficient the explosive ablation process the smaller the residual thermal injury. In other words, if a greater fraction of incident energy is used for tissue removal, a smaller fraction of incident energy is available to produce thermal injury. However, this inverse relationship may also depend on other factors such as absorption depth and tissue UTS.

Because the ablation mechanism appears to be consistent with explosive vaporization caused by metastable superheating for both wavelengths, ablation threshold should be governed only by the temperature of the tissue water. Therefore, the energy per unit volume within the tissue water at threshold should be the same for both 10.6 μm and 9.5-μm wavelengths (i.e., the threshold radiant exposure should scale with the absorption depth of water). The volumetric energy densities in the tissue water for wavelengths 10.6 μm and 9.5 μm are 690 J/cm^3 and 800 J/cm^3, respectively, and are indeed comparable.

In summary, tissue removal was consistent with explosive vaporization for both 10.6-μm and 9.5-μm wavelengths. Although the ablation mechanism appeared to be the same, 9.5-μm irradiation removed tissue more efficiently and with a smaller zone of thermal injury than at 10.6-μm irradiation. These differences may be due to the weakening of the tissue mechanical strength by directly targeting collagen. Ablation threshold scaled directly with volumentric energy density or superheat temperature within the tissue water. The observation that one can reduce or enhance thermal injury with no change in ablation mechanism by simply tuning the CO_2 laser to a different laser line may have a significant impact for a variety of dermatologic applications.

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