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# The Effect of $CO_2$ Laser Pulse Repetition Rate on Tissue Ablation Rate and Thermal Damage

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***Abstract*—The ablation rate and thermal damage in skin produced by a superpulsed $CO_2$ laser operating at pulse repetition rates between 1 and 900 Hz was measured. When delivering a fixed number of pulses (20 or 30) of equal energy, a 55–60% increase in the amount of tissue ablated was observed when the pulse repetition rate rose from 10 to 200 Hz. At pulse repetition rates greater than 200 Hz no further increase was seen. Under identical conditions, an 80% increase in the zone of thermal damage was observed when the pulse repetition rate was increased from 1 to 60 Hz. The large increases in tissue ablation and tissue damage may indicate the existence of a layer of mixed-phase (i.e., liquid and vapor) or metastable liquid which can store significant amounts of thermal energy between pulses. The data suggest that $CO_2$ lasers should be operated at relatively low repetition rates for optimal performance.**

## I. Introduction

The continuous wave $CO_2$ laser is used widely in surgical procedures because of its ability to hemostatically incise and excise tissue. Due to the significant heating involved, large zones of thermal damage in the surrounding tissue (200–1000 $\mu$m) are characteristic of the laser cutting process. Methods to reduce this thermal damage have been the subject of considerble interest for many years. One proposed method has been the use of "superpulsed" $CO_2$ lasers which have the ability to deliver a succession of short, high intensity pulses at various repetition rates [1]–[7]. However, no studies have clearly established that superpulsed lasers produce less thermal damage than continuous wave lasers. This is because the effect produced by changing a single laser parameter (e.g. pulse duration, pulse repetition rate, irradiance, etc) over a wide range of values has not been systematically investigated.

It has been suggested that at sufficiently high ablation velocities, the penetration of the resulting temperature profile into the tissue (and thus the thermal damage) is greatly reduced [8], [9]. However, in clinical practice, the ablation velocity must be slow enough to allow for control of the laser. With a continuous wave laser, it is not possible to take advantage of high ablation velocities while still allowing for manual control [9]. In the case of superpulsed lasers, high ablation velocities are theoretically possible during the pulse which may result in reduced zones of thermal damage. Furthermore, since the laser can be pulsed at a low rate, the average ablation velocity can be slow enough to allow for control by a surgeon. Thus, the effect of pulse repetition rate on the amount of tissue removed and damaged in the cutting process is of direct clinical importance. To examine these issues, we performed two sets of experiments. First, a fixed number of pulses was delivered while varying repetition rate and the amount of tissue removed was measured. In the second set of experiments, a fixed number of pulses was delivered at varying repetition rates and the width of the zone of thermal damage assessed.

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## II. Experimental Materials and Methods

### *Laser*

An RF-excited continuous wave/pulsed $CO_2$ laser (Xanar XA-50, Coherent Inc., Palo Alto, CA) was used for both experiments. Laser pulse duration was fixed at 160 $\mu$s (full width at half maximum). Pulse repetition rate could be varied over the range 0–900 Hz. Pulse energy was held constant and measured with a laser power meter (model 201, Coherent Inc., Palo Alto, CA) to be 18 $\pm$ 2 mJ. The output of the laser was focused to a spot diameter of 460 $\mu$m ($1/e^2$) by an adjustable zoom lens provided by the manufacturer. Spot size was measured by translating a knife edge across the beam and measuring the transmitted laser power with a power meter.

### *Tissue Removal Experiments*

Full-thickness pig skin with subcutaneous fat in place was obtained immediately postmortem. Samples were created with a 6 mm biopsy punch and then mounted on an analytic balance (model AE163, Mettler Instrument Corp., Highstown, NJ) so that the ablated particles would land off the weighing pan. The balance was connected to a personal computer which recorded the tissue mass at a rate of 2.4 Hz. Samples were irradiated with either 20 or 30 pulses at repetition rates between 10 and 900 Hz. The surface temperature of the samples was between 22 and 24°C prior to irradiation. Fig. 1 is a typical record of the mass of a sample undergoing laser irradiation. Mass loss before and after irradiation was due to evaporation of water from the tissue into the surrounding environment. Evaporation was corrected for by extrapolating the curve following ablation back to the time when laser pulses were first delivered. Note that although the laser pulses were delivered during a short time interval (22–3000 ms), the balance required additional time (approximately 2–5 s) to equilibriate to the change in sample mass.

### *Thermal Damage Experiments*

Full-thickness pig skin with subcutaneous fat in place was obtained immediately postmortem. The samples were then irradiated with either 20 or 30 pulses delivered at pulse repetition rates between 1 and 900 Hz. After irradiation, samples were fixed in 10% formalin, embedded in paraffin, sectioned, and stained with hematoxylin and eosin.

## III. Experimental Results

### *Tissue Removal Experiments*

Fig. 2 shows the effect of pulse repetition rate on the tissue mass ablated per pulse. Ten to fifteen samples were measured at each repetition rate. Significant (t-test, $p < 0.05$) increases in ablation rate were observed at all repetition rates greater than 10 Hz. The mass ablated continued to increase with increasing pulse repetition rate until approximately 200 Hz after which no further significant increase was seen. The mass ablated per pulse when 20 pulses were delivered was never less than with 30 pulses.

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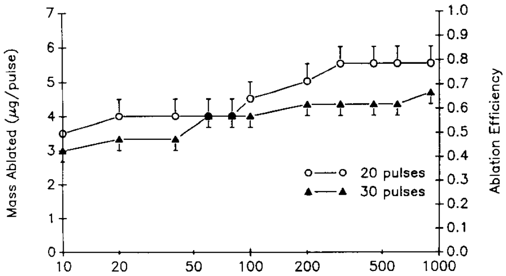


Fig. 1. Typical record of the mass of a tissue sample undergoing laser irradiation. The mass loss prior to laser irradiation was due to evaporation and corrected for as described in the test.

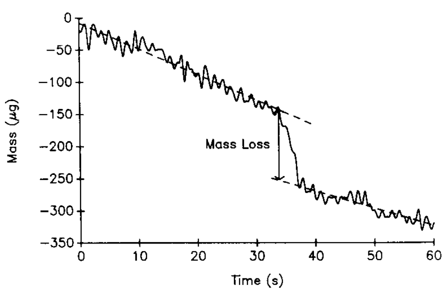


Fig. 2. Mass ablated and ablation efficiency as a function of pulse repetition rate. All increases in mass ablated are statistically significant (t-test, $p < 0.05$) except those above 200 Hz.

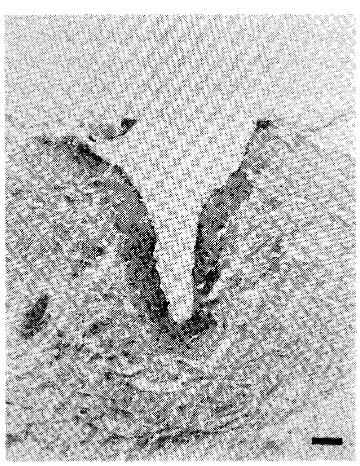

Fig. 3. Photomicrograph of a typical ablation crater produced in pig skin. 30 pulses were delivered at 600 Hz. The zone of thermal damage is approximately 90 $\mu$m. Bar = 100 $\mu$m.

### *Thermal Damage Experiments*

The zone of thermal damage was measured from the base of the ablation crater as shown in Fig. 3. Table I shows the dependence of the zone of thermal damage on pulse repetition rate. The values represent the mean of four or five samples. The zone of thermal damage increased from approximately 50 $\mu$m at repetition rates lower than 20 Hz to approximately 90 $\mu$m at repetition rates greater than 60 Hz. This difference was statistically significant (t-test, $p < 0.005$). In contrast to the tissue removal experiments, the zone of thermal damage was not significantly altered by the total number of laser pulses delivered.

TABLE I

Thermal Damage is Expressed as Mean $\pm$ One Standard Deviation. When 20 Pulses were Delivered, the Thermal Damage for Repetition Rates $\leq 20$ Hz was Statistically Different from Those $\geq 60$ Hz. When 30 Pulses were Delivered, the Thermal Damage for Repetition Rates $\leq 20$ Hz was Statistically Different from Those $> 60$ Hz. (t-test, $p < 0.005$)

| Pulse Repetition Rate (Hz) | Thermal Damage ($\mu$m) | |
|---|---|---|
| | 20 pulses | 30 pulses |
| 1 | 60 ± 10 | 50 ± 10 |
| 5 | 50 ± 10 | 50 ± 10 |
| 10 | 60 ± 10 | 60 ± 10 |
| 20 | 70 ± 10 | 70 ± 10 |
| 40 | 80 ± 10 | 70 ± 10 |
| 60 | 95 ± 10 | 75 ± 10 |
| 80 | 90 ± 15 | 95 ± 10 |
| 100 | 95 ± 10 | 100 ± 10 |
| 200 | 95 ± 10 | 85 ± 10 |
| 300 | 85 ± 10 | 100 ± 10 |
| 450 | 80 ± 10 | 100 ± 10 |
| 600 | 95 ± 10 | 100 ± 10 |
| 900 | 95 ± 10 | 90 ± 10 |

## IV. Discussion

This study demonstrates that the tissue mass ablated per pulse increases with increasing repetition rate (Fig. 2). Thus, at higher repetition rates, a larger fraction of the laser energy contributes to the tissue removal process than at lower repetition rates. The observed increases in ablation efficiency, can be explained through the hypothetical formation of a vapor-liquid layer during the ablation process as has been suggested by others [8], [10]-[14]. Because laser radiation is absorbed volumetrically in tissue, when the tissue surface reaches its phase transition temperature of 100°C and begins absorbing the latent heat, layers of tissue below the surface also reach 100°C so long as the tissue is allowed to expand isobarically. Thus, at the point when ablation actually begins, (i.e., when the tissue has fully absorbed its latent heat), a finite layer of tissue has reached 100°C and is composed of saturated vapor at the tissue surface and saturated liquid at the bottom.

Assuming that the ablation process is entirely thermal, accomplished without energy losses (e.g., heat conduction, radiation etc.), and begins at an initial tissue temperature of 22°C, we can define an ablation efficiency, $\eta$, as

$$\eta = \frac{\text{mass ablated}}{\text{mass ablated without losses}} \tag{1}$$

which allows increases in ablation efficiency to be expressed numerically. Ablation efficiencies higher than unity are possible if the mechanism of tissue removal is not exclusively thermal (e.g., explosive) as has been observed to occur in the ablation of liver using a $CO_2$ TEA laser [15].

If the vapor-liquid layer did not exist, the maximum amount of energy that could be stored for a given mass of tissue is $c(T_m - T_\infty)$ where $c$ is the specific heat of the tissue and $(T_m - T_\infty)$ is the difference between the ablation temperature and the initial tissue temperature. Without a vapor-liquid layer, the maximum percentage increase in ablation efficiency that would be expected with increasing repetition rate is the ratio between the energy required to bring a given mass of tissue to the ablative threshold temperature and the total energy required to ablate a given mass of tissue. Expressing this mathematically, we have

$$\text{maximum \% increse in } \eta \text{ (without vapor-liquid layer)} = \frac{c(T_m - T_\infty)}{c(T_m - T_\infty) + h_{fg}} \quad (2)$$

where $h_{fg}$ is the latent heat of vaporization at 1 atm. Taking the specific and latent heats of water to be $c = 4200$ J/kg°C and $h_{fg} = 2.255 \cdot 10^6$ J/kg, respectively and taking $(T_m - T_\infty) = 78$°C, we can verify that a maximum of 7 μg of tissue can be ablated by a single laser pulse and that (2) predicts a maximum increase in ablation efficiency of 13%.

This result does not agree with what we observed experimentally. With increasing repetition rate, the ablation efficiency increased from 0.50 ± 0.07 to 0.79 ± 0.07 when 20 pulses were delivered and increased from 0.43 ± 0.05 to 0.67 ± 0.05 when 30 pulses were delivered. This corresponds to increases in ablation efficiency of 60 ± 10% when 20 pulses were delivered and 56 ± 9% when 30 pulses were delivered. It may be that these large increases in ablation efficiency occur because of the existence of a vapor-liquid layer. Such a layer is capable of storing not only the energy corresponding to a temperature rise to 100°C $\{c(T_m - T_\infty)\}$ but also the latent heat $\{h_{fg}\}$. Thus, if laser pulses are delivered at a repetition rate such that the vapor-liquid layer does not have time to fully condense between pulses, a significant increase in the amount of tissue removed could occur. The characteristic time for condensation of this vapor-liquid layer is approximately 30–120 ms [16]. This condensation time suggests that increases in ablation efficiency should begin at repetition rates between ~10–30 Hz which agrees well with our experimental observations (Fig. 2). This is in contrast to earlier studies which postulated that increases in ablation rate and thermal damage would begin at repetition rates corresponding to the reciprocal of the "thermal relaxation time" [1]. This time, defined as the characteristic time for energy to diffuse across the optical penetration depth (10–20 μm for $CO_2$ laser radiation in tissue), has a value of 160–650 μs which in turn predicts enhancement in ablation efficiency at repetition rates between 1500 and 6000 Hz [1].

The existence of a vapor-liquid layer during the ablation process remains speculative. It seems unlikely that significant amounts of steam can remain in the tissue due to its large specific volume (1600 times larger than water at 1 atm). Another possibility is that as the tissue surface reaches 100°C and begins to absorb the latent heat, an insufficient number of vapor nucleation sites are formed relative to the high intensity of the laser irradiation. Furthermore, the growth of these sites may be severely compromised due to inertial and mechanical confinement by the tissue. Under these conditions, the nucleation sites may not serve as adequate temperature regulators by allowing the tissue (treated as water) temperature to exceed the saturation temperature for a given pressure. This in turn may transform the water in the tissue into a metastable thermodynamic state, being superheated to temperatures as high as 300°C even at a pressure of 1 atm [17]. However, as more nuclei form and grow, the vapor pressure at thee sites may cause the tissue to fail mechanically thereby achieving material removal. The mechanism of laser induced "explosion" has been proposed by others [18]–[20] and supported by high-speed photographs of the ablation process [21]. If explosive vaporization is indeed the actual mechanism of tissue ablation, the metastable liquid rather than the vapor-liquid layer, could represent the mechanism by which significant amounts of thermal energy could be stored between successive laser pulses.

In general, the measured ablation rates were greater when 20 pulses were delivered than when 30 pulses were delivered, independent of repetition rate. This phenomenon was probably due to the effect of varying aspect ratio. That is, when 30 pulses are delivered, a deeper ablation crater was formed which allowed the ablation products to screen and scatter the incident laser radiation and made it more difficult for the ablated tissue particles to emerge from the crater.

As shown in Table I the zone of thermal damage increased from approximately 50 μm at 1 Hz to approximately 90 μm at repetition rates greater than 40 Hz. The damage produced at the higher repetition rates was comparable to that seen when the tissue is cut by high irradiance continuous wave $CO_2$ lasers [9]. At higher repetition rates, more tissue surrounding the ablation crater is subjected to the high temperatures (≥65°C) which produce thermal damage. It may seem paradoxical that a larger zone of damage was observed at higher repetition rates since more tissue was removed at these repetition rates and less energy is available to damage the surrounding tissue. However, while there is less residual thermal energy in the tissue at the higher repetition rates, this energy is delivered over a much shorter duration of time and, as a result, this energy does not have time to diffuse away from the incision site between pulses. Because of this, energy accumulates adjacent to the incision and results in a larger zone of thermally damaged tissue. An increase in thermal damage was seen even at low repetition rates (~20–60 Hz) further indicating that the relevant tissue thermal relaxation time is on the order of tens of milliseconds rather than hundreds of microseconds as previously predicted [1].

It is unlikely that the laser can be controlled manually at repetition rates greater than about 100 Hz. This is because at a pulse repetition rate of 100 Hz, a 2 mm crater is formed in 300 ms when using the laser parameters employed in our experiments. However, at pulse repetition rates on the order of 20–60 Hz, manual control of the laser is possible (2 mm crater in 800–1500 ms) with the added benefit of smaller zones of thermal damage (≤70 μm). Thus, if minimal thermal damage and manual control is desired for clinical use, superpulsed $CO_2$ lasers should operate at relatively low repetition rates.

## V. Conclusions

The dependence of $CO_2$ laser pulse repetition rate on tissue removal and damage was examined. A substantial increase in the tissue ablated per pulse was observed between 10 and 200 Hz. An increase in thermal damage was also observed from ~50 μm at 1 Hz to ~90 μm at 60–900 Hz. We postulate that these increases are due to the existence of a vapor-liquid layer or metastable liquid which is formed during the laser ablation process. The characteristic time for condensation of the vapor-liquid layer is on the order of tens of milliseconds, consistent with the observed increase in the amount of tissue removed and damaged beginning at repetition rates as low as 20–60 Hz.

Due to high average ablation velocities, manual control of the laser at higher repetition rates is probably not possible. At lower repetition rates, a significant reduction in thermal damage is seen and manual control of the laser is possible because of reduced average ablation velocity. Thus, superpulsed $CO_2$ lasers, when operated with appropriate parameters, offer the surgeon a powerful tool for incising and excising tissue with a minimal amount of thermal damage.

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