

Hard tissue ablation with a mechanically Q-switched CO₂ laser

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ABSTRACT

Bone ablation with 400 ns pulses of a mechanically Q-switched CO₂ laser is reported. A miniature water spray was used, which alleviates tissue carbonization, even at high laser pulse repetition rates, and increases ablation efficiency. An ablation threshold of less than 2 J/cm², an optimal energy density of 10 J/cm², and a corresponding specific ablation energy of 25-30 J/mm³ was found for pig thighbone *compacta* at $\lambda = 9.57 \mu\text{m}$, and a beam waist diameter of 0.5 mm.

Keywords: CO₂ laser, Q-switch, bone, ablation

1. INTRODUCTION

Laser osteotomy has several potential advantages in comparison with a classical bone incision with mechanical saws: hemostatic and aseptic effects, intricate cut geometry etc. A reliable laser system, which emits several tens of Watt at a wavelength strongly absorbed by bone is necessary for the osteotomy. A CO₂ laser meets very well these criteria; however its application leads normally to strong charring (carbonization), which significantly delays healing¹⁻³. A hard cortical bone (*compacta*) contains on average 26 volume % of H₂O and 34 % of hydroxyapatite crystallites (the rest is collagen)⁴. The hydroxyapatite absorption coefficient α ranges at CO₂ laser wavelengths from 3500 to 5500 cm⁻¹ (Ref. 5), 4-9 times higher than that for H₂O⁶. Thus, the mineral bone component absorbs most of the laser energy and may become very hot ($T_{\text{melt}} = 1280^\circ\text{C}$ ⁷). Since the heat diffuses very quickly out of the initial absorption volume, thermal damage of surrounding tissue results and the efficiency of ablation is reduced. Using known thermal⁴ and optical^{5,6,8} constants of the tissue components we can estimate the thermal relaxation time, $\tau = 1/4\delta\alpha^2$ (δ is a thermal diffusivity). For $\lambda = 9.6 \mu\text{m}$, the τ value is only 20 μs for the *compacta*, as compared to 600 μs for a soft muscle tissue.

It has been reported, however, that very short ($\tau \leq 1 \mu\text{s}$) and intense pulses from a TEA CO₂ laser produce uncarbonized ablation craters in bone, at least when few pulses and low repetition rate of 0.5 Hz are used⁹. The very fast energy deposition in an absorption layer only a few μm thick leads to an immense pressure of evaporated bone water and collagen fragments. The resulting vapor outbreak destroys the solid structure though the temperature is far below of T_{melt} . The importance of such a water microexplosion mechanism in the ablation of bone and tooth at $\lambda \approx 3 \mu\text{m}$ (H₂O absorption maximum) has been postulated¹⁰⁻¹² and confirmed by analysis of debris particles¹³, temperature¹⁴ and micromorphology^{14, 15} of the ablation site.

In this report it is shown, that a high water content in bone tissue is a main requirement to avoid carbonization during ablation with sub- μs CO₂ laser pulses. The "wet" ablation technique¹⁶ makes high ablation efficiency feasible, while avoiding tissue carbonization even for relatively deep cuts and high laser pulse repetition rates.

2. EXPERIMENTAL

Fresh animal bone samples were irradiated *in vitro* with pulses of a mechanically Q-switched CO₂ laser¹⁷. The system is based on an industrial cw CO₂ laser, modified with a rotating chopper disk, which periodically interrupts the focal line within a conical mirror telescope (Fig. 1). Such a Q-switch is very reliable and can be used even at very high peak and average intracavity power. The resulting laser pulses are of 250 - 500 ns duration (400 ns in these experiments). The pulse repetition rate f could be adjusted by the number of the chopper slits up to several tens of kHz. The pulse energy E at $\lambda = 9.6 \mu\text{m}$ is up to 80 mJ at $f = 300 \text{ Hz}$ and up to 20 mJ at 20 kHz. The wavelength of 9.6 μm was chosen because of ~ 1.5 times higher bone absorption coefficient than at 10.6 μm .

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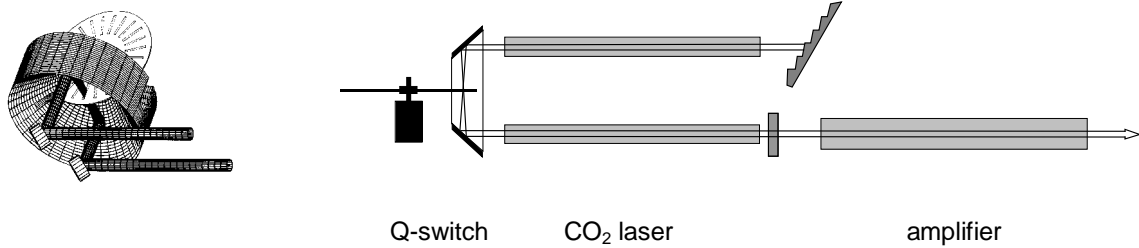


Fig. 1. Simplified schema of the CO₂ laser-amplifier system with a fast mechanical Q-switch. $\tau_{\text{pulse}} = 250 - 500 \text{ ns}$, $f = 150 - 20000 \text{ Hz}$, $E = 80 - 20 \text{ mJ}$ ($9.6 \mu\text{m}$), $M^2 = 1.1$ (nearly Gaussian beam).

A high beam quality ($M^2 = 1.1$) allows to define exactly the beam focus parameters. The beam was focused on the sample surface down to the spot diameter of $2w = 490 \mu\text{m}$ ($1/e^2$ intensity level). The corresponding focus (Rayleigh) length is 18 mm.

3. QUALITATIVE RESULTS

Initial qualitative results can be summarized as follows. Bone ablation was always accompanied with visible glowing of the ejected debris, even if the ablation crater is not thermally damaged. This indicates a partial absorption of the laser light in the debris. Only the first few laser pulses produced a clean ablation without carbonizing. When the crater depth reached several hundreds of μm , the efficiency of the ablation dropped quickly and visible thermal damage appeared even at the low repetition rate $f \approx 20 \text{ Hz}$. Prolonged irradiation and/or higher repetition rates led to a pronounced melting structure and finally to a black carbonization pattern surrounded by a white recrystallization rim. Fast laser beam scanning across the bone permitted higher f values, nevertheless clean ablation was still limited by the depth of a few hundreds of μm . An auxiliary gas jet, if applied for several tens of seconds, promoted drying of the bone tissue and only aggravates the damage.

The interpretation of these results could be as follows. Every laser pulse reduced the water content in the vicinity of the ablated volume. After some time, water content dropped to such an extent that it no longer participated in the ablation process. Subsequent laser pulses will melt and burn the solid bone components. The reduction of the ablation rate due to the parching of bone tissue has also been noticed in experiments with Er:YAG lasers ($2.94 \mu\text{m}$).^{11, 18}

4. “WET” ABLATION

In further experiments bone parching was prevented by using a miniature water spray ($\sim 2\text{-}3 \text{ ml/min}$). The application of a “cooling” water spray (usually in combination with an Er:YAG laser) is known in laser dentistry^{15, 18-21}. Despite the term, the direct cooling of the tooth tissue by the spray may be of only secondary importance. According to the mechanism discussed above, the participation of water in the ablation process would be more important. A direct liquid mediated cooling may be effective, if wavelengths are used which are relatively weakly absorbed by water (e.g. Nd:YAG laser²²).

The bone sample (a piece of hard pig thighbone *compacta* or porous pig rib *spongiosa*) was put on a rotating plate ($v = 20\text{-}55 \text{ cm/sec}$). During each turn the sample was sprayed shortly before it was exposed to the laser beam. After irradiation the sample was cut perpendicularly to the ablation groove and examined under a microscope to reveal carbonization and melting traces. The pictures of the groove profiles were taken with a CCD camera and processed using a standard image-analysis software. The groove depth D , cross-section A and effective width $w_{\text{eff}} = A/D$ were determined as a function of the equivalent pulse number N_{eq} , the energy density in the focus $\Phi = E/\pi w^2$, and the pulse overlap factor n . The equivalent pulse number shows how many laser pulses act effectively at each groove location: $N_{\text{eq}} = (\text{number of the ablation pulses}) \times (w_{\text{eff}} / \text{cut length})$. The overlap factor n is a ratio of the beam radius to the distance the sample travels between two laser pulses. Using the water spray (“wet” process) ablation of the *compacta* without carbonization was possible up to a groove depth $D \geq 10 w_{\text{eff}}$. No cracking of the bone material was observed within the range of the experimental parameters.

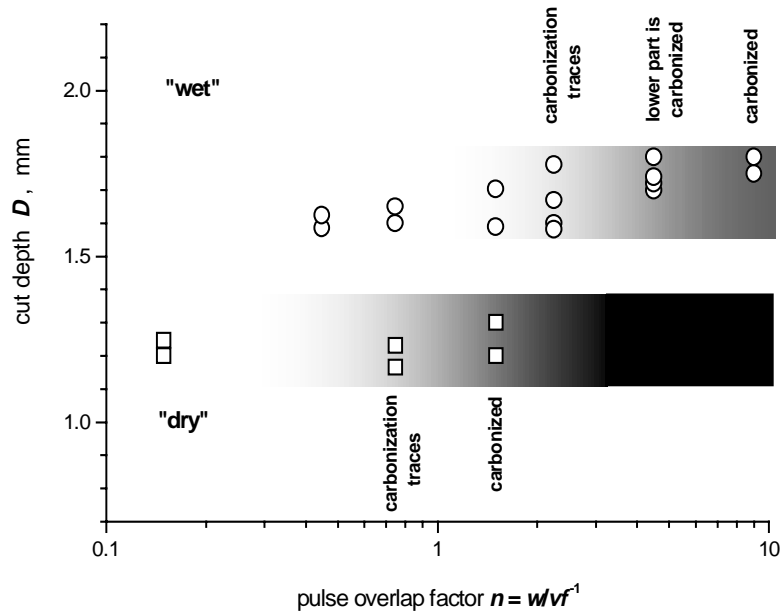


Fig. 2. Qualitative dependence of pig thighbone *compacta* carbonization on the laser pulses overlap factor n ; with (○) and without (□) water spray. $N_{eq}=280$, $\Phi=8.1 \text{ J/cm}^2$, $f=320\text{--}20000 \text{ Hz}$ (the average laser power is 5–300 W).

Figure 2 illustrates the extent of the carbonization dependent on the beam space overlap factor n . Without water spray (“dry” ablation), a massive charring is observed at $n \geq 1$. Conversely, only the lower part of a deep cut is carbonized after the “wet” ablation process, even at $n = 5$. A severe charring is observed at $n \geq 10$ in that case.

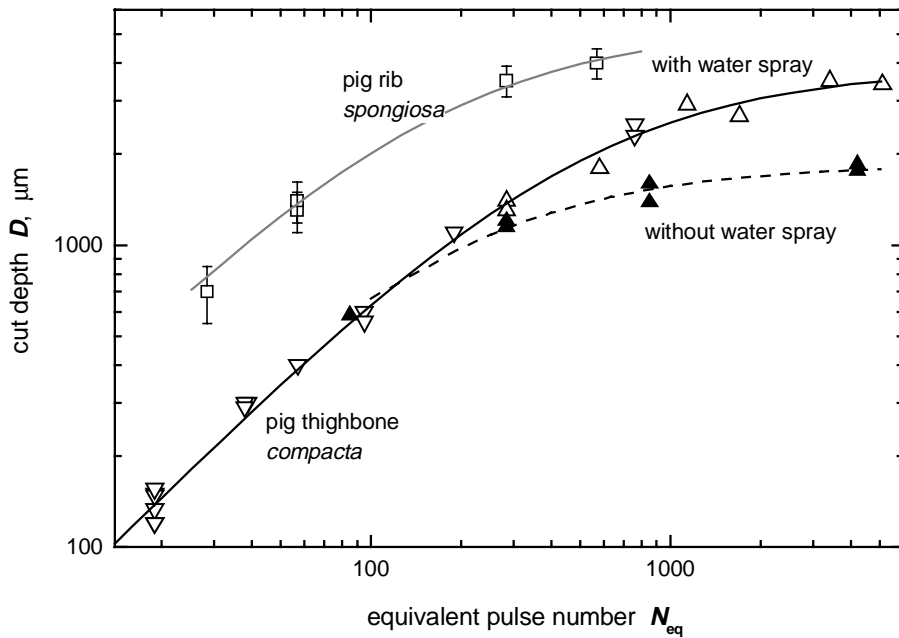


Fig. 3. Cut depth dependence on the equivalent pulse number. Pig thighbone *compacta* (▽, Δ) and pig rib *spongiosa* (□), $\Phi = 7.9 \text{ J/cm}^2$, $f = 327$ (▽), 977 (Δ, □) Hz, pulse overlap factor $n = 0.2$ (▽), 0.5 (Δ, □).

In Figure 3 cut depth is plotted against N_{eq} ; the cut depth without water spray is also shown for comparison. The difference between these two cases is obvious at high N_{eq} (deep cuts), where the “dry” ablation was less effective and is accompanied by carbonization and melting traces even at $n < 0.5$ (non-overlapped laser pulses). The measured effective cut width was $250 \pm 50 \mu\text{m}$ in both cases and seems to be independent of N_{eq} . There was no dependence found on the cut direction. The relatively large scatter of the experimental values is presumably due to variations in the tissue properties within the bone. The porous *spongiosa* is much easier to cut due to higher water content and lower mechanical strength.

The slowing-down of ablation observed at high N_{eq} can be explained by considering the geometry of the cut groove. When it is produced with a Gaussian laser beam profile, the groove is U-shaped after a few pulses, V-shaped if $D > w_{\text{eff}}$, and reveals irregularities if $D \gg w_{\text{eff}}$. The inner surface of the groove grows with depth. While that does not change the actual volume of the absorbing layer and the specific rate of energy deposition ($\text{J}/\text{mm}^3/\text{s}$), it does increase the undesirable diffusion of heat out of the absorption volume. Perhaps even more detrimental is the fact that ablation products can not escape from the deep cut quickly enough and hence absorb much of the light. At $D \gg w_{\text{eff}}$, the asymmetry of the cut also reduces the light energy reaching the lowest part of the groove. It is also conceivable that the water is unfavorably distributed within the deep groove. The tissue parching discussed above is another important factor reducing ablation efficiency at high N_{eq} without the water spray.

From the measured dependencies of D and A on N_{eq} , the single pulse ablation characteristics can be determined: an ablation depth $\delta D = dD/dN_{\text{eq}}$, an ablation volume δV and a specific ablation energy $E/\delta V$ per pulse (Fig. 4). Under the experimental conditions ($\Phi = 7.9 \text{ J}/\text{cm}^2$), δD was $7.5 \mu\text{m}/\text{pulse}$ at the beginning of the “wet” ablation, and dropped to $1 \mu\text{m}/\text{pulse}$ at $D = 2.4 \text{ mm}$ ($N_{\text{eq}} = 900$). The corresponding specific ablation energy increased from 30 up to $220 \text{ J}/\text{mm}^3$. The specific ablation energy of the “dry” process was much higher for deep cuts. $E/\delta V$ for *spongiosa* was several times smaller at identical conditions.

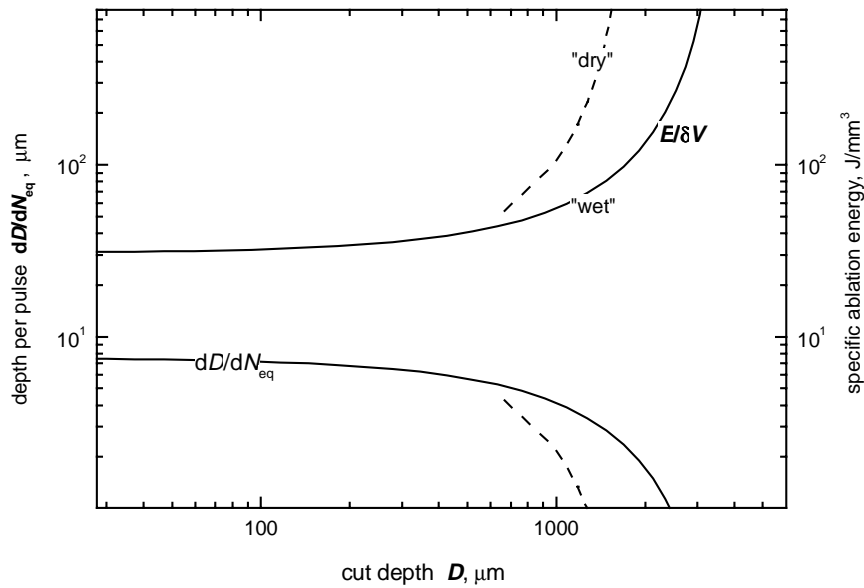


Fig. 4. Dependence of the differential laser cut depth $\delta D = dD/dN_{\text{eq}}$ and specific ablation energy $E/\delta V$ per pulse on the cut depth D (fitted curves). Pig thighbone *compacta*, experimental parameters are like in the Fig. 3.

Both δD and w_{eff} depend on the light energy density in the focus. This dependence was investigated by varying the laser pulse energy E at fixed focus diameter. The measured ablation threshold (pig thighbone *compacta*) $\Phi_{\text{th}} = E_{\text{th}}/\pi w^2$ was about $1.4 \text{ J}/\text{cm}^2$ for both “wet” and “dry” processes. This value agrees quite well with $1 \pm 0.15 \text{ J}/\text{cm}^2$ reported in Ref. 9 ($9.6 \mu\text{m}$, 900 ns pulses). Above the threshold the ablation depth initially grew quickly with Φ , but the growth rate slowed down noticeably at $\Phi > 10 \text{ J}/\text{cm}^2$. The effective cut width exhibited similar behavior. In the case of “wet” ablation, it can be described to a rough approximation by

$$w_{\text{eff}} = w\sqrt{\ln(\Phi/\Phi_{\text{th}})/2}. \quad (1)$$

Eq. (1) arises from the Gaussian light intensity distribution, $\phi(r) = 2E/\pi w^2 \cdot \exp(-2r^2/w^2)$, when supposing Φ_{th} does not change with the cut depth, neglecting the heat transfer and absorption by debris. Disagreement between the experimental values and eq. (1) is observed at high Φ_{th} , which is especially noticeable in the case of “dry” ablation (smaller w_{eff} values).

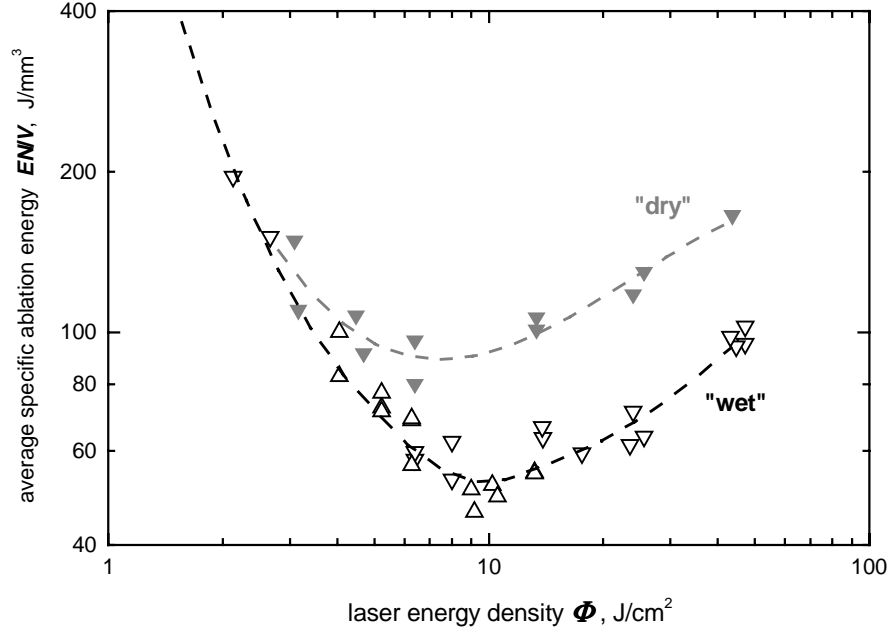


Fig. 5. Specific ablation energy as a function of laser pulse energy density Φ in the focus of the Q-switch CO₂ laser beam. Pig thighbone *compacta*, $f = 977$ (∇), 4900 (Δ) Hz, $n = 0.4$ (∇), 2.2 (Δ).

Calculated from the measured values of D and w_{eff} (or A), the average specific ablation energy (total irradiation energy/cut volume) is a strong function of Φ (Fig. 5). It has a minimum at 9-10 J/cm² for the “wet” and 7-8 J/cm² for the “dry” process. The corresponding minimal ablation energies were about 50 and 85 J/mm³, respectively. These values were averaged over the cut depth of 1.7 mm ($N_{\text{eq}} = 285$) for the “wet” and 0.9 mm ($N_{\text{eq}} = 210$) for the “dry” process. The average specific energy at the beginning of the ablation process is much smaller and obviously coincides with $E/\delta V$ at $N_{\text{eq}} \rightarrow 1$. The value of 25-30 J/mm³ was measured for the “wet” process at $\Phi = 9$ J/cm² and $N_{\text{eq}} = 20$. The existence of an optimal energy density can be explained by two competing mechanisms. The pulse intensity grows with Φ (the pulse duration is unchanged), that leads to faster energy deposition and more efficient ablation. On the other hand, the volume of ejected vapor and ablation debris also increases with Φ , it absorbing and, after ionization, reflecting and scattering more and more light.

The described mechanisms are not specific to CO₂ laser induced ablation. The saturation of the ablation depth with increasing energy density have also been observed by ablation of bone and tooth tissues with an Er:YAG laser^{3,11,13,18,20}. As an example, the specific ablation energy was recalculated from the data on ablation depths reported in Ref. 3 (Er:YAG laser, $\tau_{\text{pulse}} = 180$ μ s). If the effective crater width is approximated with eq. (1), than the resulting specific ablation energy has a minimum at 70-80 J/cm². Such a high optimal Φ value can be mainly attributed to the 400 times longer pulse and correspondingly smaller peak pulse intensity of the Er:YAG laser, as compared with the Q-switch CO₂ laser.

5. CONCLUSION

With short (400 ns) pulses of the Q-switch CO₂ laser and use of the water spray “clean” cuts (no visible carbonization) in hard *compacta* bone material have been demonstrated, even at high spatial pulse overlap factors (up to $n \approx 5$) and pulse repetition rates up to several kHz. At $E = 17$ mJ/pulse, $f = 5$ kHz and scanning velocity of 30 cm/sec clean cuts of 30-40 μm depth per pass were possible. The maximal depth is limited practically to several mm because of the reduction of the ablation efficiency with the cut depth. We found an ablation threshold of 1.4 J/cm², an optimal energy density of 9-10 J/cm², and a corresponding specific ablation energy of 25-30 J/mm³ for pig thighbone *compacta* at $\lambda = 9.57$ μm .

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