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Wet bone ablation with mechanically Q-switched high-repetition-rate CO_2 laser

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Abstract. Carbonization-free ablation of bone tissue with 400-ns pulses from a Q-switched CO_2 laser using a miniature water spray is demonstrated. An ablation threshold of $1.4\,\mathrm{J/cm^2}$, an optimal energy density of $9–10\,\mathrm{J/cm^2}$, and a corresponding specific ablation energy of $25–30\,\mathrm{J/mm^3}$ are found for pig thighbone *compacta* at $\lambda = 9.57\,\mu\mathrm{m}$ and a beam waist diameter of 0.5 mm. The water spray alleviates tissue carbonization even at high laser pulse repetition rates and increases ablation efficiency.

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CO_2 lasers are widely used for ablation of soft biological tissues. The process is based on the fast, explosive evaporation of intracellular and interstitial water. A moderate thermal effect outside the ablation zone causes a therapeutically useful sealing of small capillaries. In contrast, CO_2 laser mediated ablation of bone tissue usually leads to the formation of a charring (carbonization) layer, which significantly delays healing [1–3]. A hard cortical bone (*compacta*) contains only 12–15 wt. % of H_2O and 55-60 wt. % of hydroxyapatite crystallites (the rest is collagen) [4]. The absorption coefficient α of the pure hydroxyapatite at CO_2 laser wavelengths ranges from 3500 to $5500\,\mathrm{cm}^{-1}$ [5], which is $4-9$ times higher than that for H_2O [6]. Thus, the mineral bone component absorbs most of the laser energy and may become very hot ($T_{\mathrm{melt}} = 1280\,°\mathrm{C}$ [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 thermal [4] 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 $\lambda = 9.6\,\mu\mathrm{m}$, the τ value is only $20\,\mu\mathrm{s}$ for the *compacta*, as compared to $600\,\mu\mathrm{s}$ for a soft muscle tissue.

It has been reported, however, that very short ($\tau \leq 1\,\mu\mathrm{s}$) and intense pulses from a TEA CO_2 laser produce uncarbonized ablation craters in bone, at least when a few pulses and a low repetition rate of 0.5 Hz are used [9]. 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 although the temperature is far below T_{melt}. The importance of such a water microexplosion mechanism in the ablation of bone and tooth at $\lambda \approx 3\,\mu\mathrm{m}$ (H_2O absorption maximum) has been postulated in [10–12] and confirmed by analysis of debris particles [13], temperature [14], and micromorphology [14, 15] of the ablation site.

We will show in this report that a high water content in bone tissue is a main prerequisite for avoiding carbonization during ablation with sub-μs CO_2 laser pulses. The "wet" ablation technique we describe here makes high ablation efficiency feasible, while avoiding tissue carbonization even for relatively deep cuts and high laser pulse repetition rates.

1 Experimental arrangement and qualitative results

Fresh animal bone samples were irradiated in vitro with pulses of a mechanically Q-switched CO_2 laser [16]. The system is based on a highly reliable industrial cw CO_2 laser, modified with a rotating chopper disk, which periodically interrupts the focal line within a conical mirror telescope. The resulting laser pulses are of 400–450 ns duration (FWHM). 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 is 80 mJ at $f = 300\,\mathrm{Hz}$ and $\lambda = 9.6\,\mu\mathrm{m}$. The beam was focused on the sample surface down to a spot diameter of $2w = 490\,\mu\mathrm{m}$ ($1/e^2$ intensity level). The corresponding focus (Rayleigh) length is 18 mm.

Our initial qualitative results can be summarized as follows. The bone ablation is always accompanied by 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 produce a clean ablation without carbonizing. When the crater depth reaches several hundred μm, the efficiency of the ablation drops quickly and visible thermal damage appears even at the low repetition rate $f \approx 20\,\mathrm{Hz}$. Prolonged irradiation and/or higher repetition rates lead 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 permits higher f values, nevertheless clean ablation is still limited to a depth of a few hundred µm. An auxiliary gas jet, if applied for several tens of seconds, promotes drying of the bone tissue and only aggravates the damage.

The interpretation of these results is relatively straightforward. Every laser pulse reduces the water content in the vicinity of the ablated volume. After some time it drops to such an extent that water no longer participates 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 µm) [11, 17].

2 Wet ablation

In a second series of experiments we tried to prevent bone parching by using a miniature water spray (~ 2–3 ml/min). The application of a "cooling" water spray (usually in combination with an Er:YAG laser) is known in dentistry (see [15, 17–20] and references therein). 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 seem to be more important. A direct liquid-mediated cooling may be effective if one uses wavelengths that are relatively weakly absorbed by water (e.g. Nd:YAG laser [21]).

The bone sample (a piece of hard pig thighbone *compacta* or porous pig rib *spongiosa*) was put on a rotating plate ($v = 20$–55 cm/s). During each turn the sample was sprayed shortly before it was exposed to the laser beam. After irradiation the bone sample was cut perpendicular to the ablation groove and examined under a microscope to reveal traces of carbonization and melting. The pictures of the groove profiles were taken with a CCD camera and processed using a PC. The groove depth D, cross-section A, and effective width $w_{\mathrm{eff}} = A/D$ 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: $N_{\mathrm{eq}} =$ (number of the ablation pulses) $\times$ (w_{eff}/cut length). The overlap factor n is the 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 $n = 2-5$ and a groove depth $D \geq 10 w_{\mathrm{eff}}$. No cracking of the bone material was observed within the range of the experimental parameters.

In Fig. 1 cut depth is plotted against N_{eq}; the cut depth without water spray ("dry" ablation) is shown for comparison. The difference between these two cases is obvious at high N_{eq} (deep cuts), where the "dry" ablation is much less effective and is accompanied by carbonization and melting even at $n < 0.5$ (non-overlapping laser pulses). The measured effective cut width was 250 ± 50 µm in both cases and seems to be independent of N_{eq}. The relatively large scatter of the experimental values is presumably due to variations in the tissue properties within the bone.

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 $D > w_{\mathrm{eff}}$, and reveals irregularities if $D \gg w_{\mathrm{eff}}$. 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 (J/mm^3/s), it does increase the undesirable diffusion of heat out of the absorption volume. Perhaps even more detrimental is the fact that ablation products cannot escape from the deep cut quickly enough and hence they absorb much of the light. At $D \gg w_{\mathrm{eff}}$ 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.

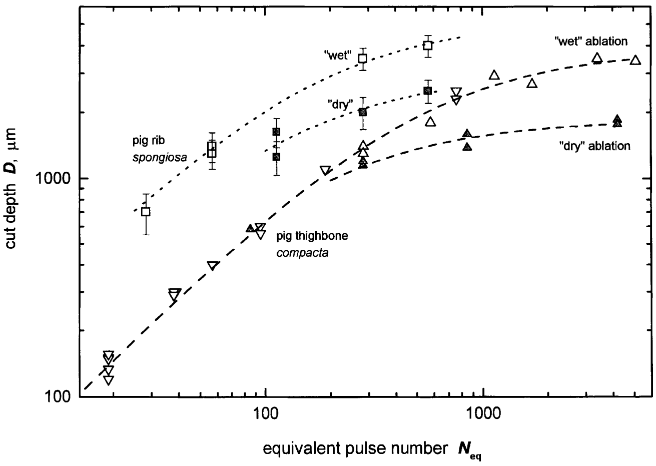

Fig. 1. Dependence of the cut depth D in pig thighbone *compacta* (∇, Δ) and pig rib *spongiosa* ($\square$) on the equivalent pulse number N_{eq} for the case of "wet" and "dry" (no water spray) ablation with 400 ns CO_2 laser pulses. Experimental conditions: $\lambda = 9.57$ µm, $\Phi = 7.9$ J/cm^2, $w = 0.25$ mm, $f = 327(\nabla)$, $977(\Delta, \square)$Hz, $n = 0.2(\nabla)$, $0.5(\Delta, \square)$

From the measured dependencies of D and A on N_{eq} we can find the single pulse ablation characteristics: an ablation depth $\delta D = \mathrm{d}D/\mathrm{d}N_{\mathrm{eq}}$, an ablation volume δV and a specific ablation energy $E/\delta V$ per pulse. Under the experimental conditions of Fig. 1 ($\Phi = 7.9$ J/cm^2), δD is 7.5 µm/pulse at the beginning of the wet ablation, and drops to 1 µm/pulse at $D = 2.4$ mm ($N_{\mathrm{eq}} = 900$). The corresponding specific ablation energy increases from 30 to 220 J/mm^3.

Both δD and w_{eff} depend on the light energy density in the focus. We investigated this dependence for pig thighbone *compacta* by varying the laser pulse energy E at fixed focus diameter and practically unchanged pulse duration. The measured ablation threshold $\Phi_{\mathrm{th}} = E_{\mathrm{th}}/\pi w^2$ is about 1.4 J/cm^2 for both wet and dry processes. This value agrees quite well with the value of 1 ± 0.15 J/cm^2 reported in [9] (9.6 µm, 900 ns pulses). Above the threshold the ablation depth initially grows quickly with Φ, but the growth slows down noticeably at $\Phi > 10$ J/cm^2. The effective cut width exhibits similar behavior. In the case of wet ablation, it can be described to a rough approximation by

$$w_{\mathrm{eff}} = w\sqrt{\ln(\Phi/\Phi_{\mathrm{th}})/2}\,, \tag{1}$$

which arises from the Gaussian light intensity distribution, $\phi(r) = 2E/\pi w^2 \exp(-2r^2/w^2)$, when supposing that Φ_{th} does not change with the cut depth, neglecting the heat

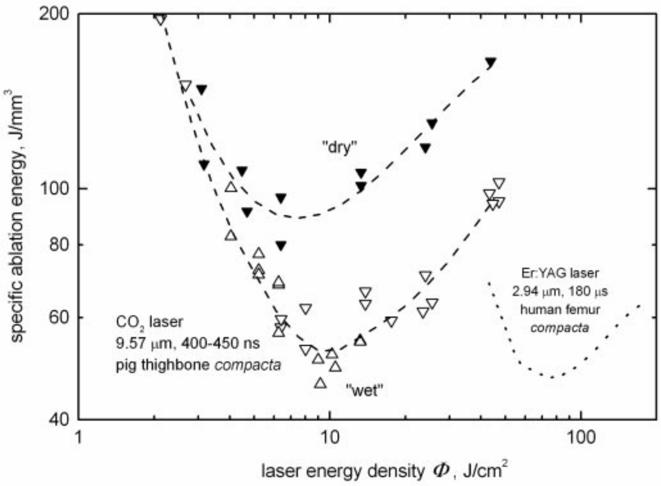

Fig. 2. Specific ablation energy as a function of laser pulse energy density Φ in the focus. The *triangles* represent our measurements with a Q-switched CO_2 laser at pig thighbone *compacta*: $\lambda = 9.57\,\mu$m, $\tau = 400–450$ ns, $w = 0.25$ mm, $f = 977(\nabla)$, $4900(\Delta)$Hz, $n = 0.4(\nabla), 2.2(\Delta)$. The *dotted curve* depicts the specific energy for an Er:YAG laser, recalculated from the experimental data of [3] (human femur *compacta*, $\lambda = 2.94\,\mu$m, $\tau = 180\,\mu$s, $f = 5$ Hz, $w = 0.3$ mm, ablation crater width is approximated with equation (1), $\Phi_{th} = 8\,J/cm^2$)

transfer and absorption by debris. Disagreement between the experimental values and equation (1) is observed at high Φ_{th}, which is especially noticeable in the case of dry ablation (smaller w_{eff} values).

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. 2). It has a minimum at 9–10 J/cm^2 for the wet and 7–8 J/cm^2 for the dry process. The corresponding minimal ablation energies are about 50 and 85 J/mm^3, respectively. These values are averaged over the cut depth of 1.7 mm ($N_{eq} = 285$) for the wet and 0.9 mm ($N_{eq} = 210$) for the dry process. The average specific energy at the start of the ablation process is much smaller and obviously coincides with $E/\delta V$ at $N_{eq} \to 1$. We measured the value of 25–30 J/mm^3 for the wet process at $\Phi = 9\,J/cm^2$ and $N_{eq} = 20$. The existence of an optimal energy density can be explained by two competing mechanisms. The pulse intensity grows with Φ, and 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, first absorbing and, after ionization, reflecting and scattering more and more light.

Such mechanisms are not specific to CO_2 laser ablation. The saturation of the ablation depth with increasing energy density or the existence of an optimal Φ value have also been reported in [3, 11, 13, 17, 19] for ablation of bone and tooth tissues with an Er:YAG laser. To illustrate this we recalculate the specific ablation energy from the data on ablation depths reported in [3] for 180 μs pulses from an Er:YAG laser. The effective crater width is approximated with (1). The resulting specific ablation energy (dotted curve in Fig. 2) has a minimum at considerably higher Φ values (70–80 J/cm^2), as compared with the Q-switch CO_2 laser. We attribute this difference mainly to the 400 times longer pulse and correspondingly smaller peak pulse intensity of the Er:YAG laser.

3 Conclusion

The bone ablation technique using short (400–500 ns) CO_2 laser pulses and water spray provides cuts with no visible carbonization even at high pulse repetition rates and spatial pulse overlap factors ($n \leq 5$). Moving the beam at a velocity of 30 cm/s we can use our Q-switch CO_2 laser ($f = 5$ kHz, $E = 17$ mJ) to produce clean cuts of 30–40 μm depth per pass in hard pig thighbone *compacta*. The maximum depth is so far limited to several mm because of the reduction of the ablation efficiency for deeper cuts. We found for the *compacta* an ablation threshold of 1.4 J/cm^2, an optimal energy density of 9–10 J/cm^2, and a corresponding specific ablation energy of 25–30 J/mm^3 at $\lambda = 9.57\,\mu$m. Preliminary experiments show that other hard biological tissues (tooth, wood) could also be ablated very cleanly using this technique.

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