First Assessment of a Carbon Monoxide Laser and a Thulium Fiber Laser for Fractional Ablation of Skin

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Background and Objectives: A recent generation of 5,500 nm wavelength carbon monoxide (CO) lasers could serve as a novel tool for applications in medicine and surgery. At this wavelength, the optical penetration depth is about three times higher than that of the 10,600 nm wavelength carbon dioxide (CO2) laser. As the amount of ablation and coagulation is strongly influenced by the wavelength, we anticipated that CO lasers would provide extended coagulation zones compared with the CO2 laser, which is possibly desirable depending on the clinical goal. The effect of deep ablation combined with additional thermal damage on dermal remodeling needs to be further confirmed with in vivo studies. Lasers Surg. Med. © 2020 The Authors. Lasers in Surgery and Medicine Published by Wiley Periodicals, Inc.

Key words: carbon dioxide laser; CO2 laser; carbon monoxide laser; CO laser; thulium fiber laser; Tm:fiber laser; coagulation; thermal damage; fractional ablation; laser skin resurfacing

INTRODUCTION

Fractional ablation of skin is characterized by microscopic channels of locally removed epidermis and dermis [1]. Ablative fractional lasers are clinically applied to treat actinic keratosis, pigmentation disorders, acne scars, and deep wrinkles [2]. The carbon dioxide (CO2) laser is one of the most widely used lasers for laser skin resurfacing of photodamaged skin [3].

Focused CO2 laser beams cause immediate evaporation of tissue with minimal thermal damage due to the strong absorption in tissue water. The depth and width of the ablation crater in combination with the amount of thermal injury principally define the clinical outcome in terms of remodelling effects and wound

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healing [4]. Accordingly, there is a demand for optimizing the exposure parameters that involve wavelength, energy per pulse, pulse duration, pulse number, and temporal pulse shape. Again, the chosen wavelength defines further parameters, such as the optical absorption coefficient and minimal focal beam diameter. This highlights the interest in novel laser sources that enable more tailored treatment opportunities [5].

Patel and Kerl [6] firstly reported on laser emission of carbon monoxide (CO) in the 5,000–5,400 nm spectral range in 1964. In contrast to numerous reports on CO2 lasers, there has been less research work done on the CO laser and its applications [7,8], probably because of unreliable CO laser technology. In the past, the CO laser power was low in output and the stability of active molecules was limited. Recently, the CO laser technology has improved and has become similar to the well-established CO2 laser [9]. A tighter focusing of the CO laser beam could be regarded as an advantage of a shorter wavelength, as the beam waist diameter scales linearly with wavelength.

The thulium fiber (Tm:fiber) laser is known as a non-ablative laser that is qualified in dermatology for the treatment of melasma and mild to moderate photodamage [10]. The Tm:fiber laser was therefore originally intended for superficial epidermal indications due to the low absorption coefficient.

Deep ablation into the dermis of more than 1.5 mm in depth with a controlled amount of coagulation is believed to cause the effects of skin tightening [11]. The ratio of ablation to coagulation is dictated by multiple parameters including the choice of wavelength. There have been many approaches, which aimed to deliver additional thermal damage to the skin while maintaining a deep ablation crater. For example, the operation of a dual-wavelength laser system that integrates an ablative and a non-ablative laser [12] or a single-wavelength laser system that applies a sequence of a high-energy short pulse for ablation and a low-energy long pulse for additional coagulation [13]. However, the beam alignment of two different wavelengths can be challenging and the application of multiple pulses can lead to plume shielding and tissue shrinkage during the irradiation. Thus, it is desirable to achieve a controlled amount of ablation and thermal damage by the use of one single pulse and one single wavelength.

In this comparison study, the ablation and coagulation capabilities of a novel CO laser and a high power Tm:fiber laser were investigated due to their recent availability. Furthermore, we discuss their suitability as fractional ablative lasers and point out the advantages and disadvantages of the CO laser. We anticipated major differences in ablation depth and thermal damage in comparison to the CO2 laser, based on the water absorption properties, as shown in Figure 1 [14]. The choice of the “ideal laser” always depends on the clinical goal that demands either ablation or coagulation or even both. The determination of optimal treatment parameters of each laser is not part of this initial ex vivo study as wound healing processes are required to evaluate the in vivo skin response to laser therapy.

Fig. 1. Tissue water is the target absorber in the skin. The pure water absorption coefficient $\mu_w$ is shown at the different laser wavelengths $\lambda$ [14]. As the water content of the skin is about 70%, the modified absorption coefficients are 630 cm$^{-1}$ at 10,600 nm, 193.2 cm$^{-1}$ at 5,500 nm, and 79.8 cm$^{-1}$ at 1,940 nm wavelength.

Materials and Methods

Laser Systems

Laser exposures on skin tissue were performed with a modified, clinical CO2 laser, a prototype CO laser, and a commercially available Tm:fiber laser. The laser specifications are summarized in Table 1. A pulse duration of 2 milliseconds and a focal spot diameter of 108–120 µm were applied. We selected 2 milliseconds to guarantee ablation well above threshold in case of the Tm:fiber laser. Differences in temporal pulse structure depicted in Figure 2 will be outlined in the following.

10,600 nm CO2 Laser. As the gold standard system, a 10,600 nm fractional CO2 laser system (UltraPulse; Lumenis Ltd., Yokneam, Israel) was applied. This clinical CO2 laser was modified. Modulation of the radio frequency power supply allowed external programming of the temporal pulse profile. The modulation of the pulse width was achieved by turning the voltage source on and off at 50 kHz repetition frequency. The very fast switching generates a pulsing structure with a more constant power level over the pulse width while maintaining the microoscillation structure. The emitted laser pulse is composed of a train of microoscillations with a repetition frequency of 2 MHz (220 nanoseconds FWHM-micropulses). We adjusted a fixed pulse width of 2 milliseconds delivered over a range of pulse energies ranging from 20 to 132 mJ. The DeepFX handpiece (UltraPulse; Lumenis Ltd., Yokneam, Israel) provided a focal spot diameter of 120 µm.

5,500 nm CO Laser. The new generation of 5,500 nm CO laser (GEM-100; Coherent Inc., Santa Clara, CA) was used, and this is a continuous wave radio frequency excited waveguide device that was gated to generate 2 milliseconds laser pulses comparable to those of the CO2 laser. We measured a 6.5 MHz repetition frequency (90 nanoseconds...
FWHM-micro pulses) within the 2 milliseconds pulses. The laser beam transport system was purged with dry nitrogen gas to avoid absorption of CO laser light in atmospheric air and water vapor. We generated a 113 μm focal beam diameter by using a biconvex CaF₂ lens (LB5774, f = 25.4 mm; Thorlabs Inc., Newton, NJ). A 1,940 nm Tm:fiber laser (TLR-120-1940; IPG Photonics Corporation, Birmingham, AL) was gated to generate 2 milliseconds pulses. In contrast to the microoscillation structure of the CO₂ and CO laser pulses, the Tm:fiber laser exhibited a rectangular-shaped temporal pulse structure with no visible microstructure. The Tm:fiber laser beam was focused to a 108 μm beam diameter using a planoconvex ZnSe lens (88-007, f = 254 mm; Edmund Optics Inc., Barrington, NJ).

**Laser Parameters**

**Pulse duration and thermal relaxation time.** For a selective and spatially confined tissue effect, a wavelength that is predominantly absorbed by water, a tightly focused 1,940 nm Tm:fiber laser. A 1,940 nm Tm:fiber laser (TLR-120-1940; IPG Photonics Corporation, Birmingham, AL) was gated to generate 2 milliseconds pulses. In contrast to the microoscillation structure of the CO₂ and CO laser pulses, the Tm:fiber laser exhibited a rectangular-shaped temporal pulse structure with no visible microstructure. The Tm:fiber laser beam was focused to a 108 μm beam diameter using a planoconvex ZnSe lens (88-007, f = 254 mm; Edmund Optics Inc., Barrington, NJ).

![Fig. 2. Temporal pulse shapes with normalized amplitudes. Originally recorded at 115 mJ pulse energy using a photoelectric infrared detector (PEM-10.6; Vigo System S.A., Ozarow Mazowiecki, Poland). The magnification visualizes microoscillations in the CO₂ and CO laser pulse structures, whereas no microstructure is visible in the Tm:fiber laser pulse. (A) Modified, custom-built CO₂ laser with pulse-width modulation. (B) CO laser. (C) Tm:fiber laser.](image-url)
beam, and a short and high-energy pulse is needed [15]. No substantial heat diffusion during irradiation takes place if the pulse duration \( \tau \) is shorter than the thermal diffusion time \( \tau_{tc} \)

\[
\tau < \tau_{tc}.
\]  

(1)

The thermal diffusion time is defined as [16]

\[
\tau_{tc} = \frac{1}{k \mu a},
\]  

(2)

where \( \mu_a \) is the absorption coefficient using 70% of the water absorption coefficients and \( k = 1.43 \times 10^{-3} \text{cm}^2/\text{s} \) is the thermal diffusivity of water [17]. Pulses of 2 milliseconds are thermally confined for the CO laser (\( \tau_{tc} = 18.7 \text{milliseconds} \)) and the Tm:fiber laser (\( \tau_{tc} = 109.4 \text{milliseconds} \)), but they are slightly longer than the thermal relaxation time for the CO\(_2\) laser (\( \tau_{rc} = 1.8 \text{milliseconds} \)).

**Temporal pulse structure measurements.** We monitored the laser pulses using a thermal sensor head (10 A; Ophir Optronics Ltd., Jerusalem, Israel) and a photoelectric infrared detector (PEM-10.6; Vigo System S.A., Ozarow Mazowiecki, Poland). The rise and fall time of the photoelectric sensor was \( \leq 0.5 \) nanoseconds (Fig. 2).

On the basis of the high repetition rates of the micropulses, single 2 milliseconds macropulses comparable to the Tm:fiber laser pulses with no pulse stacking and no blow-off effects were assumed. Also, the duration of the micropulses was longer than the stress confinement times. Stress confinement is defined for pulse durations \( \tau \) smaller than the stress confinement time \( \tau_{sc} \) [18]

\[
\tau_{sc} = \delta/c_a,
\]  

(3)

where \( c_a = 1,540 \text{m/s} \) is the longitudinal speed of sound in tissue [19] and \( \delta \) is the optical penetration depth. Hence, the stress confinement times are approximately 10.3 nanoseconds for CO\(_2\) laser and 33.6 nanoseconds for CO laser so that spallation effects were neglected.

Water is the dominant absorber at 5,500 and 10,600 nm, as collagen absorption is about 50% lower at these wavelengths. Microscale thermal confinement refers to the diameter \( d \) of dermal collagen fibrils, which have a characteristic thermal diffusion time \( \tau_{ms} \) of 69 nanoseconds according to [18]

\[
\tau_{ms} = d^2/\kappa,
\]  

(4)

where \( \kappa \) is the thermal diffusivity of collagen fibers. The micropulses of the CO\(_2\) and CO lasers, which are 220 and 90 nanoseconds, respectively, are sufficiently long to allow thermal diffusion into the dermal collagen matrix (photothermal pathway). Therefore, we anticipated that microscale thermal confinement does not influence the ablation process. Shorter laser pulses achieving a microscale thermal confinement would lead to an explosive ablation with mechanical tearing (photomechanical pathway) [20].

**Spatial beam profile measurements.** The focused beam diameters were measured using the scanned knife-edge technique [21]. The beam diameter and Rayleigh range of the lasers assuming a Gaussian beam profile are shown in Table 1.

**Laser Tissue Exposures**

With approval by the Massachusetts General Hospital Institutional Review Board, we used skin tissue samples obtained from abdominoplasty that will otherwise be discarded. No patient information was attached to the tissue samples. The tissue was carefully adjusted in the focal plane using a three-dimensional translation stage. The skin was exposed to single pulses on the epidermal side.

**Etch Depth and Thermal Damage Measurements**

After irradiation, the tissue was embedded in a frozen optimal cutting temperature medium. Nitro blue tetrazolium chloride stain enabled the demarcation between blue-stained viable cells and unstained thermally damaged cells. The stained sections were imaged using a Nanozoomer RS Digital Pathology System (Olympus America Inc., Melville, NY), which has a scanning resolution of 0.23 \( \mu \text{m/pixel} \). The NPD.view software (Hamamatsu Photonics, Hamamatsu City, Japan) was used for measuring the ablation crater dimensions.

The etch depth \( \delta_{abl} \) of the ablation crater was defined as the maximum depth of ablated tissue. The mean lateral thermal injury \( \bar{\delta}_a \) was defined as the thickness of the damaged tissue at the crater edges located at 50% of the etch depth. The mean residual thermal injury \( \bar{\delta}_b \) was defined as the thickness of damaged tissue at the bottom of the crater. The thickness of the thermal damage zones was averaged over 10 sections.

The ablation-to-coagulation-ratio (ACR) was calculated by dividing the etch depth \( \delta_{abl} \) by the mean residual thermal injury \( \bar{\delta}_b \) at the bottom of the ablation crater as follows:

\[
ACR = \frac{\delta_{abl}}{\bar{\delta}_b}.
\]  

(5)

**Ablation Models**

Basic ablation models such as the blow-off model for pulse lengths that are much shorter than the thermal relaxation time (\( \tau \ll \tau_{tc} \)) and the steady-state model for pulse lengths that are much longer than the thermal relaxation time (\( \tau \gg \tau_{tc} \)) can be used to characterize the ablation behavior and to predict the etch depth \( \delta \). The blow-off model assumes that no heat diffusion occurs during irradiation and that tissue removal occurs only after the end of the laser pulse. The complete energy deposition is required until ablation is initiated. The blow-off model predicts the etch depth according to [22]
where $\mu_a$ is the absorption coefficient, $H_0 = I_0 \tau$ is the radiant exposure, $I_0$ is the irradiance, $\tau$ is the pulse duration, and $H_{th}$ is the threshold of ablation. The ablation threshold is defined as the minimum radiant exposure required for tissue ablation to occur [23].

However, the steady-state model assumes that heat diffusion occurs during irradiation and that tissue removal occurs at the beginning of the laser pulse, which continues during the whole irradiation. The steady-state model anticipates an etch depth as follows [22]:

$$\delta = \frac{1}{\mu_a} \ln \left( \frac{H_0}{H_{th}} \right) = I_0 \tau \frac{\mu_a}{H_{th}}.$$  

(7)

However, the steady-state model and the blow-off model can be inaccurate for pulse durations $\tau$ that are close to the thermal relaxation time $\tau_{tc}$ as calculated in Equation 2. The Hibst modification connects the blow-off and the steady-state ablation models. Here, the etch depth is given by [24,25]

$$\delta = \frac{1}{\mu_a} \ln \left( \frac{H_0}{H_{th}} \right) = I_0 \tau \left( \frac{\gamma H_{th}}{H_0} - \gamma - 1 \right).$$  

(8)

Originally, Hibst defined the parameter $\gamma = \frac{\mu_a}{\mu_{a2}}$ to combine the effects of plume absorption and ablation within a single model, where $\mu_{a1}$ is the absorption coefficient of the plume and $\mu_{a2}$ is the absorption coefficient of the tissue [24]. Hence, Equation 8 is reduced to the blow-off model in Equation 6 if the plume exhibits the same absorption coefficient as the tissue or to the steady-state model in Equation 7 if the plume is transparent [22]. We used the parameter $\gamma$ as a fit parameter for our experimental results to approximate whether an ablation process is closer to the blow-off model ($\gamma \to 1$) or to the steady-state model ($\gamma \to 0$). The Hibst equation was fitted to the data using Matlab (R2016b; The MathWorks, Natick, MA).

RESULTS

Etch Depth

The depth of ablation increased with higher radiant exposures and with higher absorption coefficients, as expected if the different lasers were compared at the same incident radiant exposures (Fig. 3A). Due to the different
ablation thresholds of the lasers, a better approach was the comparison of the ablation results as a function of radiant exposure $H_0$ normalized to the ablation threshold $H_{th}$, which is $H_0/H_{th}$ (Fig. 3B). In this comparison, it was remarkable that, as expected from a theoretical point of view, the CO laser generated higher etch depths than the modified CO$_2$ laser (Fig. 4). The superficial ablation using the Tm:fiber laser was a result of the low radiant exposure compared with the high ablation threshold (Figs. 5 and 6).

**Thermal Damage**

The thermal damage zone increases with a higher optical penetration depth, which can be seen in Figures 5 and 6. As expected, increasing the pulse energy will also consequently lead to greater thermal damage zones. The largest coagulation zones were achieved by using the Tm:fiber laser. The thermal damage zones at the bottom and at the edges of the ablation crater, respectively, are plotted in Figure 7A and B as a function of radiant exposure normalized to the ablation threshold. Surprisingly,
Fig. 6. Ablation craters produced by 2 milliseconds single pulses at 184–195 mJ pulse energy. The histology sections were stained with nitro blue tetrazolium chloride (NBTC). The thermal damage is highlighted with black lines. The etch depth $\delta_{abl}$, thermal damage at the bottom $x_{\bar{b}}$, and at the edges $x_{\bar{e}}$ of the ablation crater were measured. (A) Clinical CO$_2$ laser without modifications at 184 mJ ($\delta_{abl} = 3$ mm, $x_{\bar{b}} = 0.12$ mm, and $x_{\bar{e}} = 0.10$ mm). (B) CO laser at 195 mJ ($\delta_{abl} = 1.98$ mm, $x_{\bar{b}} = 0.16$ mm, and $x_{\bar{e}} = 0.16$ mm). (C) Tm:fiber laser at 194 mJ ($\delta_{abl} = 0.35$ mm, $x_{\bar{b}} = 0.23$ mm, and $x_{\bar{e}} = 0.24$ mm).

Fig. 7. Thermal damage of the ablation craters plotted against the radiant exposure $H_0$ normalized to the ablation threshold $H_{th}$ for the modified CO$_2$ laser, the CO laser, and the Tm:fiber laser. (A) Thermal damage at the bottom of the ablation crater $x_{\bar{b}}$. (B) Thermal damage at the ablation crater edges $x_{\bar{e}}$ at 50% of the etch depth. (C) Ablation-to-coagulation-ratio (Equation 5).
Ablation Threshold was 3.8 times higher for both, the CO2 and the Tm:fiber laser in comparison with 70% of the water ablation threshold (15.3 J/cm²). Similar to our results two times deeper ablation craters than the CO2 laser if the applied radiant exposure is well above the ablation threshold.

In these experiments, the Hibst model could not be used to describe the relationship between ablation depth and radiant exposure. The Tm:fiber laser ablation curve (γ = 0.756) showed a logarithmic dependence on radiant exposure, as expected, that was fitted using the blow-off model (γ → 1), which assumes ablation to take place at the end of the laser pulse [22]. Figure 3C shows a good linear fit on the semi-logarithmic scale, which proves the blow-off model. However, the steady-state model (γ → 0) was not applicable for the CO₂ laser and CO laser results even though the γ-values were close to zero (0.03 and 0.13) for the CO₂ laser and the CO laser, respectively. The etch depth did not increase linearly with radiant exposure, as was expected for a steady-state ablation process. The steady-state model assumes that ablation occurs with the beginning of the laser pulse and continues during the laser pulse [22]. The limiting factors of the ablation depth could be the Rayleigh range (Table 1), shielding effects by ejected materials, tissue shrinkage, and other non-linear effects. Similar findings were reported by Evers et al. [5].

### Thermal Damage

If the pulse is shorter than the thermal relaxation time, the thermal damage zone will be primarily determined by the optical penetration depth [15]. A lower absorption coefficient is inversely related to higher optical penetration depth, meaning that the thermal damage zone will be larger. Additional thermal damage can be considered as beneficial or harmful depending on the clinical goals.

One of the first reports on pulsed CO₂ laser ablation of tissue by Walsh et al. [15] have shown the effect of pulse duration on the thermal damage zone. At 2 milliseconds pulse duration, a thermal damage zone of up to 170 μm was measured. This higher coagulation zone was an effect of multiple pulses (>50). In contrast, we delivered 2 milliseconds single pulses that resulted in thermal damage zones of up to 120 μm. The least thermal damage was produced by the CO₂ laser and the most pronounced thermal damage was observed for the Tm:fiber laser. Interestingly, at radiant exposures that were 10 and 40 times higher than the ablation threshold, the CO laser exhibited about two times larger coagulation zones than the CO₂ laser. This was an effect of the three times lower absorbing wavelength of the CO laser. Surprisingly, the thermal damage zone did not significantly increase with increasing radiant exposure (Fig. 7A and B).

Unexpectedly, the thermal damage at the bottom and at the walls of the ablation crater was very similar in thickness. As we expected more thermal damage at the crater bottom, this could also be an effect of hot steam formation that transmitted heat into the crater walls by the leaving gas [5,15].

Ablative lasers that operate in the region of the water absorption peak are the 2,940 nm Erbium-doped yttrium
aluminium garnet (Er:YAG) laser and the 2,790 nm Erbium-doped yttrium-scandium-gallium-garnet (Er:YSGG) laser, which have an absorption coefficient in skin of about 8,033 and 2,513 cm⁻¹, respectively [14]. At these wavelengths, the absorption is about 13–41 times higher compared to the CO laser. Precise tissue ablation of up to 200 μm in depth was reported at 100 J/cm² radiant exposure using the Er:YSGG laser [27]. The Er:YSGG laser generates minimal thermal damage of 8–25 μm. Thermal damage zones of only 3–7 μm were reported for the Er:YAG laser [15,28]. The Er:YSGG and Er:YAG lasers will generate much smaller etch depths than the CO laser when compared at radiant exposures normalized to the ablation threshold. Besides the lower tissue penetration, the high water absorption leads to a pure ablation leaving minimal thermal damage, which would reduce the effect of skin tightening.

**Exposure Parameters**

The micropulses of the CO₂ laser and the CO laser contribute to higher peak powers, resulting in superior ablation when compared with the flat-shaped pulse of the Tm:fiber laser [29]. The CO₂ laser had a 137.5 W peak power that was 2.5 times higher than its 55 W average power, while the CO laser showed a 86 W peak power that was 1.4 times higher than its 60 W average power. Recent “superpulsed” Tm:fiber lasers achieve peak powers of 500 W, pulse rates of up to 2 kHz, and pulse lengths of 200 microseconds to 12 milliseconds [30]. “Ultrapulsed” CO₂ lasers that are used for fractional ablation have peak powers of about 240 W, 2 MHz repetition frequency, and pulse lengths of 90 microseconds to 2 milliseconds. Thus, deeper ablations produced by the CO₂ laser and the Tm:fiber laser could be achieved by modulating the temporal pulse structure in terms of “superpulses” with higher peak powers in combination with smaller spot sizes. Further, it would be interesting to investigate the relationship between modulating the temporal pulse structure and thermal damage formation.

**Comparison of CO and CO₂ Laser Beam Delivery**

The rigidity of the articulated mirror arm that is integrated into clinical CO₂ laser systems is a substantial disadvantage for surgical use. Flexible optical waveguides anticipated for the CO laser would facilitate handling [7]. Novel fiber technology for CO₂ laser achieved lower loss levels and higher damage threshold, however, the power delivery capacity is still limited to 10–20 W average power [31,32]. Despite the low output powers at the end of the fibers, the ablation threshold will be exceeded. However, the fiber delivery of 100 W CO₂ and CO lasers for fractional ablation is not yet technically attainable. In addition, the CO laser beam transport system requires an external nitrogen gas purge to avoid absorption of CO laser light in atmospheric air by water vapor as the water absorption in the gas phase is more than 10 times greater at 5,500 nm compared with 10,600 nm wavelength, which makes the handling more inconvenient. The clinical CO₂ laser system is already equipped with a continuous dry purge air to the laser system.

**Ablation Metrics**

The Hibst fit was not reliable as the fitted absorption coefficients of skin were about three times lower for the CO₂ laser and the CO laser, and almost two times lower for the Tm:fiber laser in comparison with 70% of the water absorption coefficients. The tissue ablation threshold determined from the experiments was almost four times higher for both, the CO₂ laser and the CO laser, and it was 15 times higher for the Tm:fiber laser when compared with the ablation threshold of water. The ablation threshold of skin is higher, as the ultimate tensile strength that results from the rigid collagen network of the skin needs to be overcome. The skin is a heterogeneous material consisting of different skin layers, such as stratum corneum, epidermis, and dermis, where the higher Young’s modulus of the stratum corneum (1,998 MPa) and the epidermis (102 MPa) results in higher material stiffness compared to the dermis (10.2 MPa) [33].

**Ablation-to-Coagulation-Ratio (ACR)**

The ACR could serve as a relevant parameter in the adjustment of the amount of thermal edge effects, where more coagulation is desired in the treatment outcome. An ablative laser would be characterized by a high ACR, as ablation is more dominant than coagulation. However, a coagulative laser would be characterized by a low ACR, where coagulation is more dominant than ablation. The maximum of the ACR results from the limited increase of etch depth in relation to thermal damage zone, which could be an effect caused by the Rayleigh range.

**Potential of the Tm:fiber Laser**

The Tm:fiber laser is generally known to be a non-ablative laser [10]. Due to the high ablation threshold, only superficial tissue removal with large thermal damage zones was observed. At 100 W laser power, a maximum etch depth of 380 μm and a coagulation zone of 300 μm were achieved using 2 milliseconds pulses. Tissue studies in urology have shown similar ablation and coagulation results for the incision of bladder neck tissue using a 40 W Tm:fiber laser, 10 milliseconds pulse, and 200 mJ pulse energy. Ablation and coagulation of about 400–500 μm were observed [34]. In the previous study by Fried and Murray, a spot diameter of 1.1 mm and longer pulses were used. Although the etch depth and coagulation zone at the crater bottom were similar to our results, the width of ablation including thermal damage zones is about four times larger. For dermatological indications, a higher precision of ablation is required, which could be obtained with the use of smaller spot sizes. Higher etch depths should also be obtainable by the use of smaller spot diameters (enabled by the higher Rayleigh range) and higher pulse energies (enabled by longer pulses considering the longer thermal relaxation time).
CONCLUSION

With the availability of a novel 100 W CO laser, we were interested in comparing the ablation and coagulation capabilities of the CO laser to the gold standard CO2 laser. At similar exposure parameters and at the same multiple of the ablation threshold, the CO laser was capable of generating higher etch depths with about two times larger thermal damage zones. In this configuration, the Tm:ﬁber laser generated superﬁcial craters with massive thermal damage. Higher etch depths are expected by the use of smaller spot diameters and/or higher pulse energies.

Further adjustments such as a reduction of spot size are desirable. For the same spot size, laser sources with a shorter wavelength would provide a higher Rayleigh range (longer depth of focus) [9]. Also, smaller ablation diameters in the skin could allow faster wound healing. Although ex vivo studies are great for the analysis of ablation depth and thermal damage, in vivo studies are required to show whether additional thermal damage can enhance the skin rejuvenation results without increasing adverse effects.

All in all, advantages of the CO laser with a shorter wavelength and lower absorption coefﬁcient in contrast to the CO2 laser could be the generation of smaller spot sizes, longer working distances, increased thermal damage zones, and more convenient implementation of fiber delivery [7,9]. However, the operation of the CO laser is currently constrained by the requirement of nitrogen gas purging. The experimental results indicate that CO and Tm:ﬁber lasers can be used for ablation of skin with increased amounts of coagulation. The additional coagulation could potentially enhance skin tightening effects of the fractional laser treatment due to collagen shrinkage, which needs to be veriﬁed in the next step [11].

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