Journal of Lasers in Medical Sciences

J Lasers Med Sci 2026;17:e9

Original Article

http://journals.sbmu.ac.ir/jlms

doi:10.34172/jlms.2026.09

Ablation Depth and Surface Morphology of Dentin After Irradiation with a Tunable TEA-CO_2 Laser in the Wavelength Range of 9.3 – 10.6 µm

Hadiseh Ghalkhanbaz[1], **Maryam Ilchi Ghazaani**[2*], **Batool Sajad**[1]

[1]Department of Atomic and Molecular Physics, Faculty of Physics, Alzahra University, Tehran, Iran
[2]Photonics and Quantum Technologies Research Institute, Nuclear Sciences and Technologies Research Institute (NSTRI), Tehran, Iran

***Correspondence to**
Maryam Ilchi Ghazaani,
Email: milchi@aeoi.org.ir

Received: September 1, 2025
Accepted: December 21, 2025
ePublished: May 3, 2026

Abstract

Introduction: This study evaluated morphological and structural changes in extracted human dentin following irradiation with a pulsed tunable TEA-CO_2 laser (9–11 µm) using scanning electron microscopy (SEM). The ablation rate was quantitatively determined at different wavelengths and pulse numbers while assessing preservation of dentinal tubules.

Methods: Nearly flat dentin specimens were irradiated at four wavelengths (9.3, 9.6, 10.3, and 10.6 µm) with a constant energy density of 7.36 J/cm², pulse width of 200 ns, and repetition rate of 1 Hz. Samples were grouped by cooling method (air or water) and exposed to 1, 5, 10, 20, or 30 pulses. Surface morphology was examined by optical microscopy and SEM. Single-pulse ablation depth and ablation rate (µm/pulse) were measured using a profilometer under sub-threshold thermal damage conditions. Dentinal tubule occlusion after single-shot irradiation was compared with untreated dentin controls.

Results: Air-cooled samples exhibited thermal damage, including carbonization and cracking. SEM analysis showed that water cooling led to greater thermal damage at high pulse numbers, particularly at wavelengths near 9 µm, corresponding to strong hydroxyapatite absorption and a lower ablation threshold. Up to 10 pulses, no cracks or peripheral thermal damage were observed at any wavelength. For single-pulse irradiation, the greatest ablation depth occurred at 10.3 µm with complete tubule occlusion, followed by 9.6 µm. At five pulses, maximum ablation depth was again observed at 10.3 µm (complete occlusion) and 9.3 µm (partial occlusion).

Conclusion: These findings indicate that appropriate selection of wavelength, pulse number, and cooling method enables controlled dentin removal using a pulsed TEA-CO_2 laser without peripheral tissue damage. Wavelength-dependent absorption and cooling conditions are critical for optimizing ablation efficiency while preserving dentinal tubule integrity.

Keywords: Tunable TEA-CO_2 laser, Human dentin ablation, Air-water-cooling, Thermal damage, Morphological study

Introduction

Lasers are versatile tools for dental hard tissue applications, where they are employed for caries prevention and removal, scaling, aesthetic procedures, bleaching, restorative removal, management of dentin hypersensitivity, etching, growth modulation, enamel demineralization, and diagnosis.[1-7]

Currently, the most commonly used lasers for hard tissue are from the Er: YAG family, which operate in the mid-infrared wavelength range, and pulsed CO_2 lasers, which have limited use.[8-13] Transversely excited atmospheric-pressure (TEA) CO_2 lasers have demonstrated superior efficiency, performance, and cost-effectiveness.[14–16] Additionally, high-repetition-rate TEA-CO_2 lasers can rapidly ablate hard tissues.[17–19] Under typical operating conditions, the emission pulses of TEA-CO_2 lasers exhibit a characteristic temporal structure consisting of two distinct phases: an initial high-energy spike at a duration of 100–200 ns, followed by a lower-intensity tail extending for microseconds.

In early research, CO_2 lasers were primarily used at a wavelength of 10.6 µm in continuous-wave (CW) mode, which limited their clinical adoption due to thermal tissue damage and undesirable structural changes, including extensive cracking, scaling, char formation, melting, and recrystallization in enamel, dentin, and bone.[20–26]

Subsequently, to control adverse thermal effects on the pulp and prevent heat transfer to adjacent tissues, researchers investigated the potential of CO_2 laser-based systems operating at discrete wavelengths within the 9–11 µm range.[27–32] Laboratory studies have demonstrated that optimal adjustment of laser parameters to target

Please cite this article as follows: Ghalkhanbaz H, Ilchi Ghazaani M, Sajad B. Ablation depth and surface morphology of dentin after irradiation with a tunable TEA-CO2 laser in the wavelength range of 9.3 – 10.6 µm. J Lasers Med Sci 2026;17:e9.:doi:10.34172/jlms.2026.09

tissue characteristics and the use of Q-switched (QSW) CO_2 lasers with pulse widths approximating the thermal relaxation time (TRT) of the tissue significantly improve laser ablation performance while minimizing thermal damage.[33]

The TRT in enamel and dentin varies with tissue type and specific wavelength. When the laser pulse width is shorter than the TRT, both the volume and efficiency of tissue ablation per pulse decrease due to plasma shielding.[34] Conversely, longer laser pulses are likely to create larger areas of peripheral thermal and mechanical damage.[35–37] Such peripheral thermal damage can result in cracking, thermal stress, accumulation of non-apatitic calcium phosphate (CaP) phases on the surface, and excessive collagen matrix degradation.[38]

Several studies have demonstrated that pulsed CO_2 lasers with pulse widths up to 100 µs can effectively remove dental hard tissues with minimal heat deposition within the tooth structure.[39,40] Recent surveys using CO_2 laser pulses with radiofrequency (RF) excitation and pulse durations of less than one millisecond have shown effective removal of dental hard tissues.[41] Employing water cooling (an appropriate cooling method) can also enhance the thermal management during laser procedures.

Hard-tissue ablation processes using TEA-CO_2 lasers generally involve photothermal mechanisms, ranging from precise photoablation to thermal ablation. The ablation threshold depends on the incident light wavelength and the absorption coefficient of the target tissue. Higher absorption coefficients result in greater laser energy absorption at the tissue surface, leading to more efficient ablation with reduced penetration depth.

In general, among the different CO_2 laser wavelengths, dentin exhibits the highest absorption coefficients at 9.3 µm and 9.6 µm due to the strong absorption by the mineral component hydroxyapatite (HAP). These wavelengths are particularly suitable for controlled ablation of dental hard tissues.[42] The focused beams of these laser lines create cavities with high depth-to-width ratios, enabling precise tissue removal.[43]

Surface morphology of enamel and dentin is typically studied using near-infrared real-time imaging, SEM, and confocal laser scanning microscopy (CLSM).[44–53] The significant differences in surface modifications on dental hard tissues produced by different lasers are due to variations in their absorption coefficients in the target tissue.[54–56]

In this study, the effect of laser wavelength, pulse number, and cooling method on laser dentin ablation was investigated using optical imaging and SEM analysis. Throughout all experiments, energy density (fluence), pulse repetition rate, and pulse width remained constant. Additionally, by controlling heat accumulation during laser ablation, dentinal tubule occlusion was evaluated and compared under different conditions.

Materials and Methods

Experimental Setup

Here, a TEA-CO_2 laser operating in the wavelength range of 9-11 µm was used for vertical sample irradiation. This tunable laser generates an initial high-energy spike with a duration of 200 ns at the beginning of each, followed by a longer tail lasting approximately 4 µs, which corresponds to the TRT of dentin. The laser output beam, operating at the minimum available device energy of 1 J, was directed through a beam conditioning system consisting of a 5×5 cm^2 square plexiglas plate with a central two mm-diameter circular aperture. The transmitted beam was then collimated using a flat mirror and a copper concave mirror ($f = 6$ cm), and then focused onto the target sample using a 10 cm focal-length lens. Sample positioning was maintained using a metal holder base. The schematic illustration for CO_2 laser ablation is shown in Figure 1.

As mentioned, the tunable TEA-CO_2 laser (Model 840, wavelength range 9-11 µm, pulse width 200 ns, maximum energy 5 J, average power 50 W at 10 Hz, peak power 50 MW) was used for laser irradiation. Wavelength calibration was performed using a CO_2 spectrum analyzer (Model 16A, Macken Instruments Inc., Santa Rosa, CA, USA). Laser energy was measured with a digital joule meter (Lambda Physik). Surface morphological changes were analyzed using SEM (Model T330A, JEOL Ltd., Tokyo, Japan). Optical microscopy was used for initial surface examination, and ablation depth measurements were performed using a profilometer.

Sample Preparation

Caries-free extracted human molars were collected from a dental clinic following informed patient consent. To achieve smooth, uniform surfaces, the samples were sectioned longitudinally using a precision diamond blade with computer numerical control (CNC) cutting, as shown in Figure 2. After surface preparation by sanding, the samples were mounted in orthodontic acrylic blocks for enhanced stability and control of positioning. To prevent bacterial growth, the samples were disinfected with a 0.1% thymol solution and subsequently stored in distilled water to maintain tissue hydration throughout the experimental period. To facilitate identification of the ablation zone during laser irradiation and to clearly

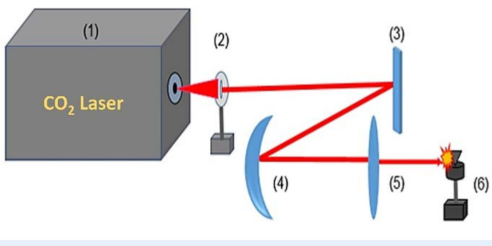


Figure 1. Schematic illustration for irradiating a dental sample including arrays of (1) pulsed TEA-CO_2 tunable laser, (2) plexiglass plate with aperture, (3) flat mirror, (4) copper concave mirror, (5) lens, and (6) metal base holding the sample

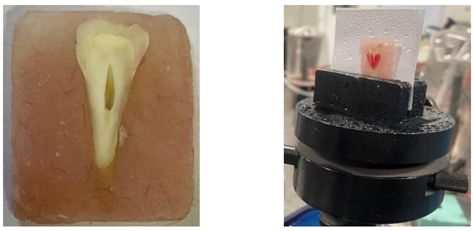
distinguish dentin from enamel regions, sample surfaces were marked with a red permanent marker. Given that the absorption of different wavelengths of the CO_2 laser is only in water and hydroxyapatite (not in melanin and hemoglobin), this work does not have a significant effect on the ablation results of the laser-tissue interaction.

Experimental Procedure

First, eight dentin samples were irradiated with a pulsed tunable TEA-CO_2 laser, pumped by a high-voltage electric discharge, at 9.3 µm, 9.6 µm, 10.3 µm, and 10.6 µm in the vertical irradiation mode. Wavelength accuracy was ensured using the back grating of the laser in conjunction with spectrometer calibration for enhanced reliability.

To examine the effect of cooling methods, the samples were divided into two groups: four samples tested with water spray cooling and the others tested with ambient air cooling. This approach allowed the evaluation of optimal cooling effects on interaction efficiency, temperature reduction, and minimization of thermal damage, including compositional and structural changes due to tissue dehydration.

The proper selection of irradiation parameters is highly effective for achieving the desired tissue ablation while minimizing thermal damage from energy accumulation. The minimum laser energy delivered to the target site was approximately 28 mJ per pulse. After passing through the optical system, the beam had a spot diameter of 0.7 mm and a cross-sectional area of 3.8×10^{-5} cm^2. The laser fluence was maintained constant at 7.36 J/cm^2 across all four wavelengths, within the ablation threshold energy density range. Each dentin sample was irradiated with 1, 5, 10, 20, and 30 pulses at a repetition rate of 1 Hz. The experiment variables throughout all stages were wavelength, pulse number, and cooling method (air versus water).



(a) (b)

Figure 2. (a) Tooth specimen prepared to access dentin and (b) marking of the dentin area with a red marker

Results

The experimental analysis was conducted in three phases. First, optical microscope images were examined to assess surface changes, followed by a detailed morphological evaluation using SEM spectroscopy. In the second phase, laser ablation depths were compared across different wavelengths. Finally, the effect of the laser pulse number on dentinal tubule occlusion was evaluated.

Optical Microscopic and SEM Investigations

Figure 3 demonstrates the effects of irradiation on stained dentin samples under ambient air cooling compared to water spray cooling at each tested wavelength. The numbers indicated in the figure represent the number of pulses applied at each ablation site. Optical microscopy images at low magnification reveal melting and carbonization of healthy dentin samples when ablated under ambient air cooling conditions. The results showed that airflow cooling was insufficient for the high power density of this laser. In contrast, when water spray cooling was employed, thermal damage was observed only at higher pulse counts, which depended on the wavelength of the incident beam.

Due to the significant thermal damage (melting and carbonization) observed in all air-cooled samples, the subsequent morphological analysis was conducted exclusively on water-cooled specimens.

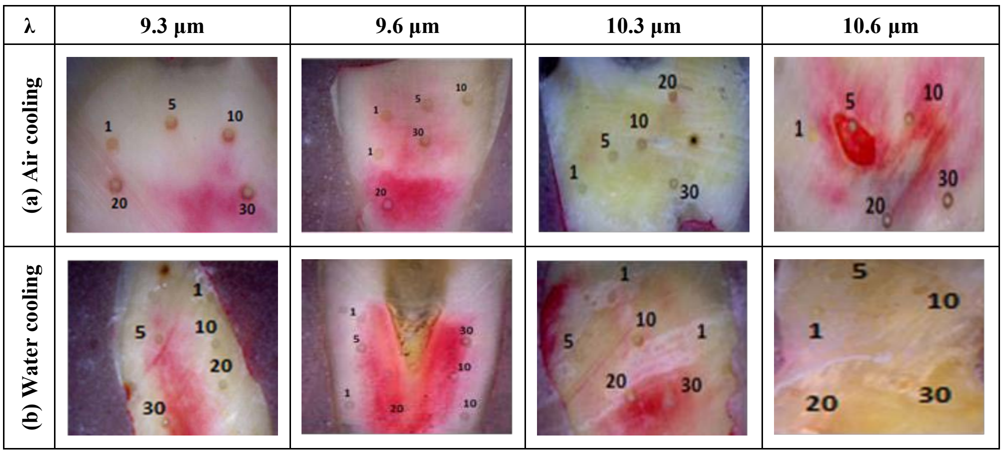


Figure 3. Optical microscopy images of laser processed dentin at wavelengths of 9.3 µm, 9.6 µm, 10.3 µm and 10.6 µm with different pulse numbers, with (a) ambient air cooling and (b) water spray (the numbers on the figure are the number of pulses applied at each point and the dentin surface of the samples is colored with a marker).

SEM images of dentin samples exposed at 9.3 µm with different pulse counts are shown in Figure 4. According to the figure, satisfactory ablation was achieved in dentin up to 10 pulses, but after that, the dentin cracked. Therefore, due to the high absorption of HAP at 9.3 µm, acceptable ablation without destructive thermal effects occurred at lower pulse counts.

In the following experiments, the laser was set to a wavelength of 9.6 µm and an energy density of 7.36 J/cm^2. In Figure 5, SEM images of irradiated dentin with water spray show that the ablation obtained with 1, 10, and 20 laser pulses is desirable, but at 30 pulses, cracking due to increased temperature and heat accumulation is observed.

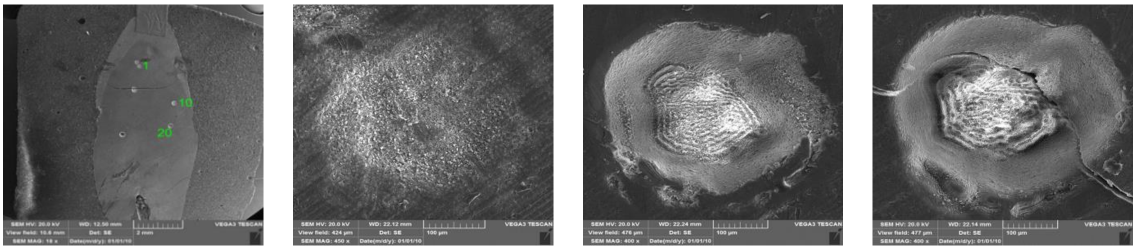
For this experimental series, laser exposure was performed at a 10.3 µm wavelength with an energy density of 7.36 J/cm^2 under water-spray cooling conditions. As demonstrated in Figure 6, dentin ablation proceeded effectively up to 20 pulses without significant thermal damage.

Then, irradiation at a 10.6 µm wavelength was performed under water spray cooling conditions. The ablated dentin shown in Figure 7 displays acceptable tissue removal without thermal damage, even at 30 pulses. It represents a significant improvement over previous experiments at other wavelengths, where effective dentin ablation under water spray cooling was limited to a maximum of 20 pulses before thermal effects became apparent.

The saturation of the ablation efficiency at higher pulse counts for other wavelengths can be attributed to plasma shielding formation. To prevent plasma formation that limits the ablation rate per pulse, either the energy fraction in the initial spike must be significantly reduced, or the pulse width must be shorter than TRT.

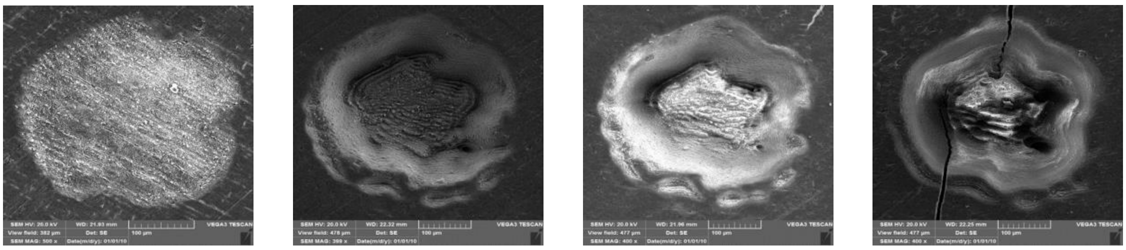
Assessment of the Ablation Depth

Figure 8 demonstrates the profound impact of the cooling method on dentin ablation patterns, as shown by



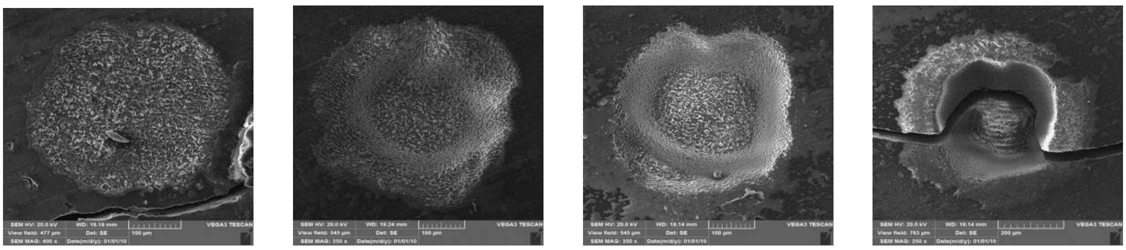
(a) (b) (c) (d)

Figure 4. SEM images of dentin irradiated at a wavelength of 9.3 µm with different pulse numbers and water spray cooling, (a) overall view of exposed dentin, (b) single-pulse, (c) 10-pulse, and (d) 20-pulse



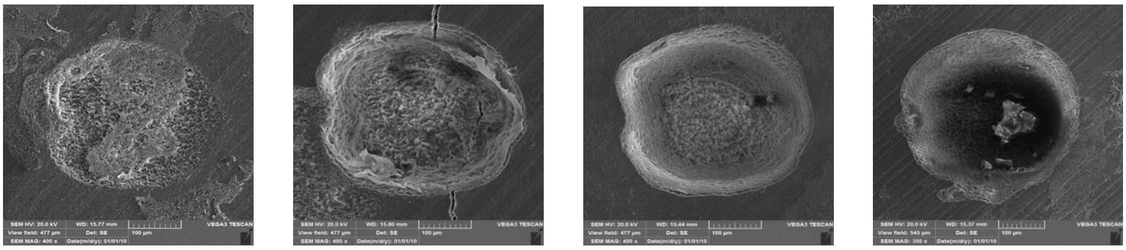
(a) (b) (c) (d)

Figure 5. SEM images of dentin exposed at a wavelength of 9.6 µm with different pulse numbers and water spray cooling, (a) single-pulse, (b) 10-pulse, (c) 20-pulse, (d) 30-pulse



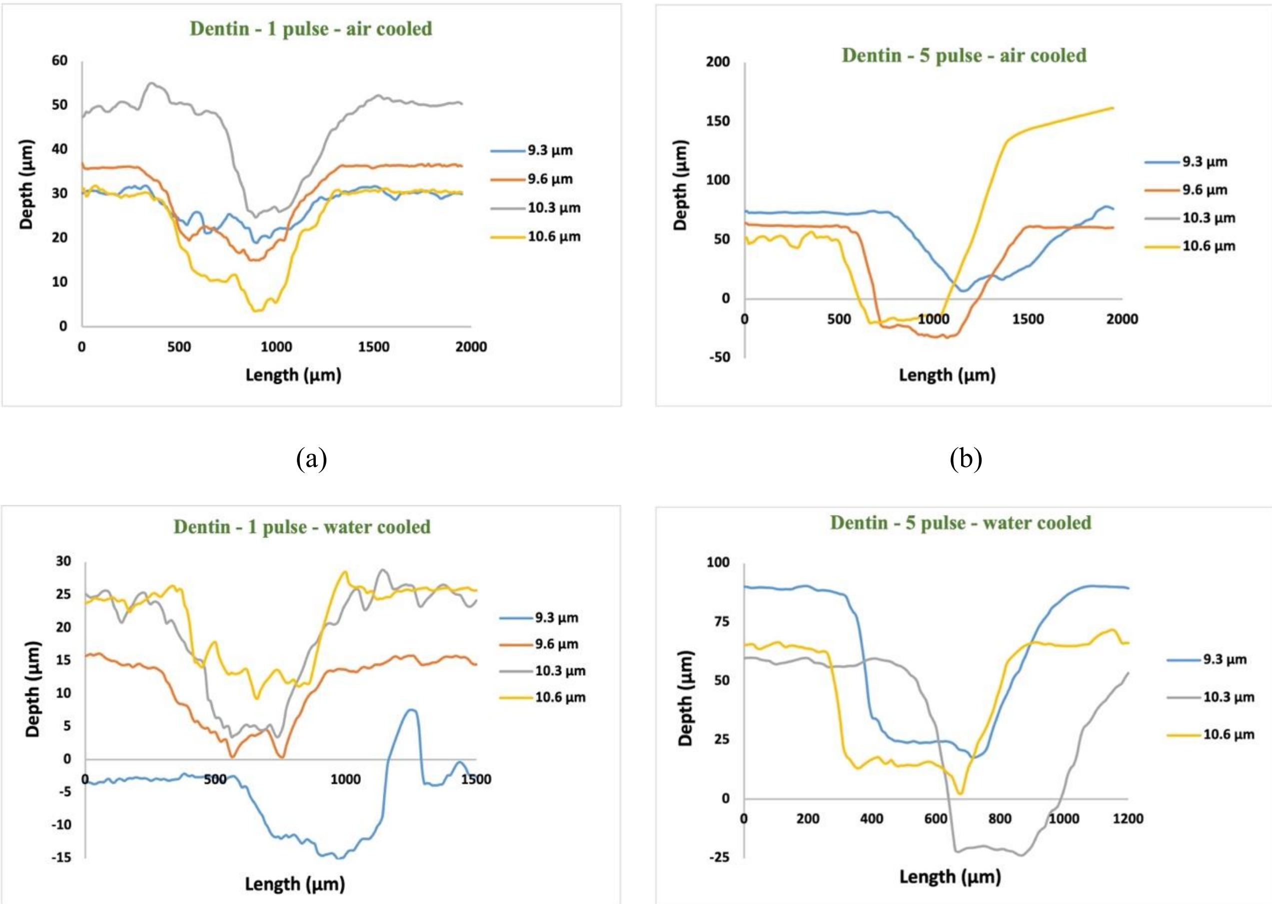
(a) (b) (c) (d)

Figure 6. SEM images of dentin irradiated at a wavelength of 10.3 µm with different pulse numbers and water spray cooling, (a) single-pulse, (b) 10-pulse, (c) 20-pulse, (d) 30-pulse



(a) (b) (c) (d)

Figure 7. SEM images of dentin irradiated at a wavelength of 10.6 μm with different pulse numbers and water spray cooling, (a) single-pulse, (b) 10-pulse, (c) 20-pulse, (d) 30-pulse



(a) (b)

(c) (d)

Figure 8. Pattern of ablation depth measured by a profilometer on dentin with ambient air-cooling for (a) single-pulse **treatments** and (b) 5-pulse, as well as water-cooling for (c) single-pulse and (d) 5-pulse at different wavelengths of the CO_2 laser

profilometer measurements across different wavelengths and pulse numbers. Under air-cooling conditions, single-pulse irradiation (panel a) produced minimal surface modification with slight positive depth values, indicating limited material removal and some surface buildup. However, the 5-pulse air-cooled treatment (panel b) resulted in dramatic positive depth measurements up to 150 μm, particularly at 10.6 μm, demonstrating extensive material accumulation and surface elevation rather than controlled ablation. Material accumulation refers to the redeposition and buildup of ablated debris on the dentin surface under air-cooling conditions.

In stark contrast, water-cooling conditions enabled accurate tissue removal, as evidenced by negative depth values throughout both single-pulse (panel c) and 5-pulse (panel d) treatments. Single-pulse water-cooled ablation achieved controlled crater formation, with maximum depths of approximately 22 μm at a 10.3 μm wavelength, which increased to 87 μm in 5-pulse ablation. Moreover, 5-pulse treatment significantly increased ablation depth to 72 μm at a 9.3 μm wavelength.

Here, negative depth values indicate measurements below the original surface baseline, indicating crater formation where material has been removed by ablation. Conversely, positive depth values indicate surface elevation above the baseline due to material accumulation. The profilometer measures vertical displacement relative to the untreated surface, with negative values representing successful material removal.

The ablation depth of dentin samples in which water was also sprayed onto the test sample during irradiation for single-pulse and 5-pulse lasers at different wavelengths is indicated in Figure 9. The lowest depth of ablation was obtained at a wavelength of 9.3 μm for a single-pulse laser and at a wavelength of 10.6 μm for a 5-pulse laser, while the ablation depth for both cases was maximum at a constant wavelength of 10.3 μm. Furthermore, the depth of ablation increased with increasing the number of laser pulses.

In addition, the results for dentin ablation depths at different CO_2 laser wavelengths are summarized in Table 1. By carefully examining the results, the highest ablation depth in both cases was achieved at 10.3 μm, with an ablation rate of 17.4 μm/pulse.

Evaluation of Dentinal Tubules

Maintaining dentinal tubule patency after laser exposure is crucial for dentin regeneration and for the transmission of sensory signals to the pulp, supporting tooth vitality. SEM analysis of irradiated samples reveals wavelength-dependent effects on tubule occlusion patterns. At wavelengths of 9.3 μm and 10.6 μm, dentinal tubules exhibit minimal occlusion and remain largely patent compared to untreated natural dentin. However, irradiation at 10.3 μm wavelengths results in significant tubule occlusion, potentially compromising dentin permeability and biological function. The morphological characteristics of natural dentinal tubules and the effects of laser irradiation at different wavelengths are illustrated in Figure 10, demonstrating the wavelength-specific impact on tubule integrity and patency following laser ablation.

Table 1. Summary of the ablation depths at different wavelengths and pulse numbers with water cooling

Wavelength (μm)	ablation depth (μm) for 1-pulse	ablation depth per pulse (μm/pulse) for 5-pulse
9.3	12.5	14.4
9.6	15.5	13.2
10.3	22	17.4
10.6	14.5	12.6

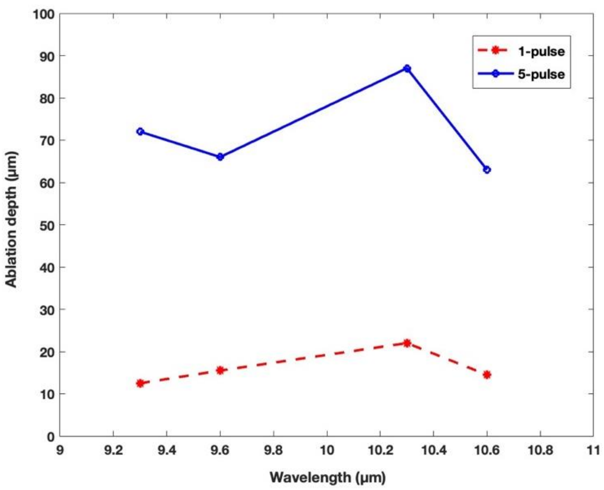


Figure 9. Depth of ablation at different wavelengths of CO_2 laser for one pulse and five pulses on dentin with water spray cooling

Discussion

As established in previous studies, dentin exhibits maximum absorption coefficients at 9.3 μm and 9.6 μm wavelengths due to strong absorption by HAP mineral (10,43). Therefore, with constant energy density (7.36 J/cm^2) and pulse width (200 ns) across all experiments, thermal damage effects become apparent at these wavelengths even at lower pulse counts. Consequently, both the ablation and thermal damage thresholds are reduced at these highly absorbed wavelengths.

Moreover, the profilometry results for the estimation of ablation depth at different wavelengths prove that water cooling is essential for achieving controlled, predictable tissue removal, while air cooling results in uncontrolled surface modification characterized by

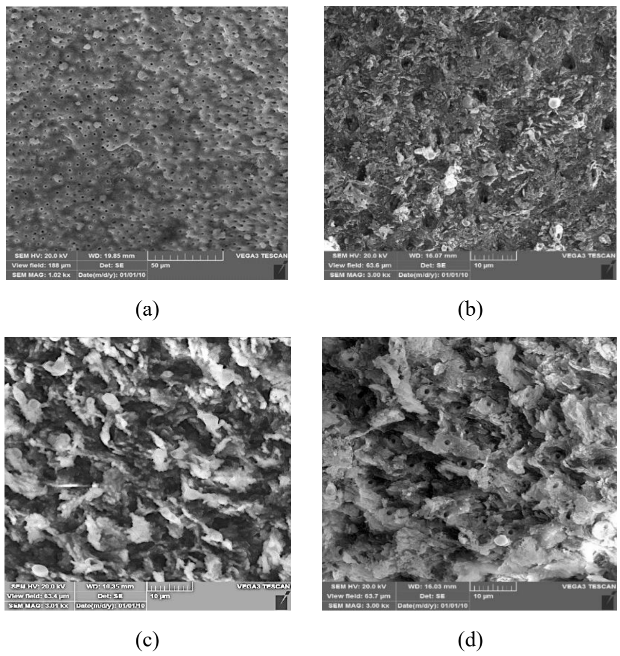


Figure 10. SEM images of dentinal tubules (a) naturally and after irradiation with wavelengths of (b) 9.3 μm, (c) 10.3 μm, and (d) 10.6 μm at a magnification of 3 kx

material redistribution rather than precise ablation. In comparison, although water spray cooling can reduce thermal effects, it significantly decreases the ablation depth and the ablation rate due to the high absorption coefficient of CO_2 lasers in water.

It can be seen that the reduction of the ablation rate at higher pulse shots is probably due to plasma shielding as well as the interaction of the laser with particles detached from the tissue.

Conclusion

The appropriate selection of laser parameters, particularly wavelength and energy density, is crucial for effective laser-tissue interaction in dental applications. Insufficient energy doses fail to produce adequate ablation, while excessively high doses risk thermal damage to surrounding tissues and potential pulp injury. Additionally, high pulse counts may lead to plasma shielding, which can limit ablation efficiency.

This study optimized TEA-CO_2 laser parameters for dentin ablation using a constant energy density of 7.36 J/cm² and a 200 ns initial pulse spike, across wavelengths from 9.3 to 10.6 µm at a 1 Hz repetition rate. Profilometry and microscopic analysis revealed that the cooling method fundamentally determines ablation quality. Air cooling led to material accumulation and surface buildup rather than controlled tissue removal, while water spray cooling enabled precise crater formation with actual tissue ablation.

Wavelength-dependent absorption characteristics significantly influenced ablation efficiency. The 9.3 µm wavelength achieved maximum ablation depth due to strong hydroxyapatite (HAP) absorption under water cooling. The 10.6 µm wavelength demonstrated superior thermal management, maintaining controlled ablation even at 30 pulses. Dentinal tubule evaluation revealed wavelength-specific effects: 9.3 µm and 10.6 µm maintained tubule patency, while 10.3 µm caused significant tubule occlusion.

The optimal parameters for controlled dentin ablation combine a 9.3 µm wavelength with water spray cooling, providing maximum ablation efficiency while preserving tubule integrity. Although water cooling may reduce absolute ablation rates due to thermal dissipation, it enables controlled, precise tissue removal, which is essential for clinical applications. These findings demonstrate that tunable TEA-CO_2 lasers with appropriate cooling systems offer significant potential for dental hard-tissue ablation when parameters are carefully optimized for the target tissue.

Authors' Contribution

Conceptualization: Maryam Ilchi Ghazaani, Batool Sajad
Data curation: Hadiseh Ghalkhanbaz, Maryam Ilchi Ghazaani
Formal analysis: Maryam Ilchi Ghazaani, Batool Sajad
Funding acquisition: Maryam Ilchi Ghazaani, Batool Sajad
Investigation: Hadiseh Ghalkhanbaz, Maryam Ilchi Ghazaani, Batool Sajad
Methodology: Maryam Ilchi Ghazaani, Batool Sajad
Project administration: Maryam Ilchi Ghazaani, Batool Sajad
Resources: Hadiseh Ghalkhanbaz, Maryam Ilchi Ghazaani
Supervision: Maryam Ilchi Ghazaani, Batool Sajad
Validation: Maryam Ilchi Ghazaani, Batool Sajad
Visualization: Hadiseh Ghalkhanbaz, Maryam Ilchi Ghazaani, Batool Sajad
Writing–original draft: Maryam Ilchi Ghazaani, Batool Sajad
Writing–review & editing: Maryam Ilchi Ghazaani, Batool Sajad

Competing Interests

The authors declare no conflict of interest.

Ethical Approval

No ethics for review study.

Funding

No funding was received for this study.

References

1. Melcer J, Farcy JC, Hellas G, Badiane M. Preparation of Cavities using a TEA CO_2 laser. In: 3rd International Congress on Lasers in Dentistry. 1992. p. Abstract #58.
2. Müllejans R, Eyrich G, Raab WH, Frentzen M. Cavity preparation using a superpulsed 9.6-microm CO2 laser--a histological investigation. Lasers Surg Med 2002;30(5):331–6. doi:10.1002/lsm.10063
3. Fried D. Laser processing of dental hard tissues (Invited Paper). In: Photon Processing in Microelectronics and Photonics IV. 2005. p. 259.
4. Marraccini TM, Bachmann L, Wigdor HA, Walsh Jr JT, Turbino ML, Stabholtz A, et al. Enamel and dentin irradiation with 9.6 µm CO2 and 2.94 µm Er:YAG lasers: bond strength evaluation. Laser Physics Letters 2006;3(2):96. doi:10.1002/lapl.200510058
5. Chan KH, Fried NM, Fried D. Selective Ablation of Carious Lesions using an Integrated Near-IR Imaging System and a Novel 9.3-µm CO(2) Laser. Proc SPIE Int Soc Opt Eng 2018;10473. doi:10.1117/12.2296025
6. Zhang OL, Yin IX, Yu OY, Luk K, Niu JY, Chu CH. Advanced Lasers and Their Applications in Dentistry. Dent J (Basel) 2025;13(1):37. doi:10.3390/dj13010037
7. Romanos GE, Estrin NE. Laser applications in peri-implantitis therapy: A narrative review. J Dent 2025;162:106081. doi:10.1016/j.jdent.2025.106081
8. Featherstone JD, Nelson DG. Laser effects on dental hard tissues. Adv Dent Res 1987;1(1):21–6. doi:10.1177/08959374870010010701
9. Fried D, Glena RE, Featherstone JD, Seka WD. Multiple-pulse irradiation of dental hard tissues at CO2 laser wavelengths. Lasers in Dentistry 1995;2394:41.
10. Featherstone JDB, Fried D. Fundamental Interactions of Laserswith Dental Hard Tissues. Medical Laser Application 2001;16(3):181–94. doi:10.1078/1615-1615-00022
11. Fantarella D, Kotlow L. The 9.3-µm CO2 dental laser: technical development and early clinical experiences. J Laser Dent. 2014;22(1):10-27.
12. Anton y Otero CI, Di Bella E, Krejci I, Bortolotto T. Effect of 9.3 µm CO2 and 2.94 µm Er:YAG Laser vs. Bur Preparations on Marginal Adaptation in Enamel and Dentin of Mixed Class V Cavities Restored With Different Restorative Systems. Frontiers in Dental Medicine 2021;2. doi:10.3389/fdmed.2021.668056
13. Roi A, Roi CI, Luca R, Munteanu RI, Miron MI. Advanced Laser Applications for Dental Caries Management. In:

Zabokova Bilbilova E, editor. Dental Caries - From Prevention to Restoration. London: IntechOpen; 2025. doi:10.5772/intechopen.1012234
14. Fried D, Murray MW, Featherstone JDB, Akrivou M, Dickenson KM, Duhn CW, et al. Dental hard tissue modification and removal using sealed TEA lasers operating at λ=9.6 and 10.6 µm. In: Lasers in Dentistry V. 1999. p. 196.
15. Luk K, Zhao IS, Gutknecht N, Chu CH. Use of carbon dioxide lasers in dentistry. Lasers in Dental Science 2019;3(1):1–9. doi:10.1007/s41547-018-0047-y
16. Hameed S, Abdul-Wahid S. CO_2 laser applications for sustainable healthcare: Reducing environmental impact in dental enamel cutting. E3S Web Conf 2025;614:04024.
17. Fan K, Fried D. A high repetition rate TEA CO_2 laser operating at λ=9.3-µm for the rapid and conservative ablation and modification of dental hard tissues. In: Lasers in Dentistry XII. 2006. p. 61370H.
18. Sattayut S, Hortong K, Kitichaiwan C. The Ablation Properties of CO(2) Laser Irradiating to Absorption Media: An In Vitro Study. Int J Dent 2012;2012:230967. doi:10.1155/2012/230967
19. Florczak-Matyjek A, Nikodem A, Kensy J, Matys J, Grzech-Leśniak K. The Efficacy of Erbium-Ion, Diode, and CO2 Lasers in Debonding Attachments Used During Overlay Orthodontic Treatment and the Risk of Hard Tooth Tissue Damage Compared to Traditional Methods—An In Vitro Study. Photonics [Internet]. 2025; 12(6):[621 p.].
20. Stern RH. Laser beam effect on dental hard tissues. J Dent Res 1964;43:873.
21. Lobene RR, Bhussry BR, Fine S. Interaction of carbon dioxide laser radiation with enamel and dentin. J Dent Res 1968;47(2):311–7. doi:10.1177/00220345680470021901
22. Kantola S. Laser-induced effects on tooth structure. VII. X-ray diffraction study of dentine exposed to a CO2 laser. Acta Odontol Scand 1973;31(6):381–6. doi:10.3109/00016357309002525
23. Melcer J. Latest treatment in dentistry by means of the CO2 laser beam. Lasers Surg Med 1986;6(4):396–8. doi:10.1002/lsm.1900060409
24. Neiburger EJ, Miserendino L. Pulp chamber warming due to CO2 laser exposure. N Y State Dent J 1988;54(3):25–7.
25. Ferreira JM, Palamara J, Phakey PP, Rachinger WA, Orams HJ. Effects of continuous-wave CO2 laser on the ultrastructure of human dental enamel. Arch Oral Biol 1989;34(7):551–62. doi:10.1016/0003-9969(89)90094-0
26. Khosrovi PM, Pogrel MA, Goodis HE, Leighty SM. Thermal effects of the carbon dioxide laser on dentin. Lasers Life Sci 1991;4(3):179–89.
27. Fried D, Glena RE, Featherstone JD, Seka W. Permanent and transient changes in the reflectance of CO2 laser-irradiated dental hard tissues at lambda = 9.3, 9.6, 10.3, and 10.6 microns and at fluences of 1-20 J/cm2. Lasers Surg Med 1997;20(1):22–31. doi:10.1002/(sici)1096-9101(1997)20:1<22::aid-lsm4>3.0.co;2-0
28. Malmström HS, McCormack SM, Fried D, Featherstone JD. Effect of CO2 laser on pulpal temperature and surface morphology: an in vitro study. J Dent 2001;29(8):521–9. doi:10.1016/s0300-5712(01)00028-8
29. Assa S, Meyer S, Fried D. Ablation of dental hard tissues with a microsecond pulsed carbon dioxide laser operating at 9.3-µm with an integrated scanner. In: Lasers in Dentistry XIV. 2008.
30. Staninec M, Darling CL, Goodis HE, Pierre D, Cox DP, Fan K, et al. Pulpal effects of enamel ablation with a microsecond pulsed lambda = 9.3-microm CO2 laser. Lasers Surg Med 2009;41(4):256–63. doi:10.1002/lsm.20748
31. Pandarathodiyil A K, Anil S. Lasers and their Applications in the Dental Practice. Int J Dentistry Oral Sci 2020;7(11):936-943. doi:10.19070/2377-8075-20000185
32. Ertl T, Muller G. Hard tissue ablation with pulsed CO_2 lasers. In: Proc SPIE 1880. 1993. p. 176–181.
33. Stanislawki M, Meister J, Mitra T, Ivanenko MM, Zanger K, Hering P. Hard tissue ablation with a free running Er:YAG and a Q-switched CO_2 laser: A comparative study. In: Applied Physics B: Lasers and Optics. 2001.
34. Lukac M, Hocevar F, Cencic S, Nemes KN, Keller U, Hibst R, et al. Effects of pulsed CO_2 and Er:YAG lasers on enamel and dentin. In: Lasers in Orthopedic, Dental, and Veterinary Medicine II. 1993. p. 169.
35. Fried D, Borzillary SF, McCormack SM, Glena RE, Featherstone JDB, Seka WD. Thermal effects on CO_2 laser-irradiated dental enamel at 9.3, 9.6, 10.3, and 10.6 microns. In: Laser Surgery: Advanced Characterization, Therapeutics, and Systems IV. 1994. p. 319–28.
36. Daniel F, Wolf DS, Richard EG, John DBF. Thermal response of hard dental tissues to 9- through 11- um CO2-laser irradiation. Optical Engineering 1996;35(7):1976–84. doi:10.1117/1.600774
37. Bell PW, Fan K, Jones RS, Fried D. Analysis of peripheral thermal damage during the rapid ablation of dentin and bone using a λ= 9.3-µm TEA CO_2 laser. In: Lasers in Dentistry XII. 2006.
38. Dela Rosa A, Sarma AV, Le CQ, Jones RS, Fried D. Peripheral thermal and mechanical damage to dentin with microsecond and sub-microsecond 9.6 microm, 2.79 microm, and 0.355 microm laser pulses. Lasers Surg Med 2004;35(3):214–28. doi:10.1002/lsm.20090
39. Koort HJ, Frentzen M. The effect of TEA-CO_2-Laser on dentine. In: 3rd International Congress on Lasers in Dentistry. 1992. p. Abstract #64.
40. Fried D, Ragadio J, Champion A. Residual heat deposition in dental enamel during IR laser ablation at 2.79, 2.94, 9.6, and 10.6 microm. Lasers Surg Med 2001;29(3):221–9. doi:10.1002/lsm.1111
41. Krapchev VB, Rabii CD, James AH. Novel CO_2 laser system for hard tissue ablation. Advanced Characterization, Therapeutics, and Systems IV. International Society for Optics and Photonics, Laser Surgery 1994;341–349.
42. Xue VW, Zhao IS, Yin IX, Niu JY, Lo ECM, Chu CH. Effects of 9,300 nm Carbon Dioxide Laser on Dental Hard Tissue: A Concise Review. Clin Cosmet Investig Dent 2021;13:155–61. doi:10.2147/ccide.S304273
43. Marraccini TM, Bachmann L, Wigdor HA, Walsh Jr JT, Stabholtz A, Zezell DM. Morphological evaluation of enamel and dentin irradiated with 9.6 µm CO2 and 2.94 µm Er:YAG lasers. Laser Physics Letters 2005;2(11):551–5. doi:10.1002/lapl.200510040
44. Takahashi K, Kimura Y, Matsumoto K. Morphological and atomic analytical changes after CO2 laser irradiation emitted at 9.3 microns on human dental hard tissues. J Clin Laser Med Surg 1998;16(3):167–73. doi:10.1089/clm.1998.16.167
45. Kimura Y, Wilder-Smith P, Krasieva TB, Liaw LH, Matsumoto K. Effects of CO_2 laser on human dentin: A confocal laser scanning microscopic study. Lasers Life Sci 1998;8(1):1–12.
46. Kimura Y, Takahashi-Sakai K, Wilder-Smith P, Krasieva TB, Liaw LH, Matsumoto K. Morphological study of the effects of CO2 laser emitted at 9.3 microm on human dentin. J Clin Laser Med Surg 2000;18(4):197–202. doi:10.1089/10445470050144047
47. Türkmen C, Günday M, Karaçorlu M, Başaran B. Effect of CO2, Nd:YAG, and ArF excimer lasers on dentin morphology and pulp chamber temperature: an in vitro study. J Endod 2000;26(11):644–8. doi:10.1097/00004770-200011000-00003
48. Watari F. Compositional and morphological imaging of CO2 laser irradiated human teeth by low vacuum SEM,

confocal laser scanning microscopy and atomic force microscopy. J Mater Sci Mater Med 2001;12(3):189–94. doi:10.1023/a:1008913828931

49. Darling CL, Fried D. Real-time near-IR imaging of laser-ablation crater evolution in dental enamel. In: Lasers in Dentistry XIII. 2007.
50. Darling CL, Fried D. Real-time near IR (1310 nm) imaging of CO2 laser ablation of enamel. Opt Express 2008;16(4):2685–93. doi:10.1364/oe.16.002685
51. Nguyen D, Chang K, Hedayatollahnajafi S, Staninec M, Chan K, Lee R, et al. High-speed scanning ablation of dental hard tissues with a λ = 9.3 μm CO2 laser: adhesion, mechanical strength, heat accumulation, and peripheral thermal damage. J Biomed Opt 2011;16(7):071410. doi:10.1117/1.3603996
52. Nelson DG, Wefel JS, Jongebloed WL, Featherstone JD. Morphology, histology and crystallography of human dental enamel treated with pulsed low-energy infrared laser radiation. Caries Res 1987;21(5):411–26. doi:10.1159/000261047
53. McCormack SM, Fried D, Featherstone JD, Glena RE, Seka W. Scanning electron microscope observations of CO2 laser effects on dental enamel. J Dent Res 1995;74(10):1702–8. doi :10.1177/00220345950740101201
54. Fried D, Ragadio J, Akrivou M, Featherstone JD, Murray MW, Dickenson KM. Dental hard tissue modification and removal using sealed transverse excited atmospheric-pressure lasers operating at lambda=9.6 and 10.6 microm. J Biomed Opt 2001;6(2):231–8. doi:10.1117/1.1344192
55. Fried D, Zuerlein MJ, Le CQ, Featherstone JD. Thermal and chemical modification of dentin by 9-11-microm CO2 laser pulses of 5-100-micros duration. Lasers Surg Med 2002;31(4):275–82. doi:10.1002/lsm.10100
56. Xue J, Wu L, Shi X, Huang J, Liang H, Zhang X. Ablation Characteristics of Hard Tooth Tissues Irradiated by 9.3 μm CO2 Laser. Zhongguo Jiguang/Chinese Journal of Lasers 2022;49(15):1507105. doi:10.3788/CJL202249.1507105