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ORIGINAL ARTICLE

Method to control depth error when ablating human dentin with numerically controlled picosecond laser: a preliminary study

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Abstract A three-axis numerically controlled picosecond laser was used to ablate dentin to investigate the quantitative relationships among the number of additive pulse layers in two-dimensional scans starting from the focal plane, step size along the normal of the focal plane (focal plane normal), and ablation depth error. A method to control the ablation depth error, suitable to control stepping along the focal plane normal, was preliminarily established. Twenty-four freshly removed mandibular first molars were cut transversely along the long axis of the crown and prepared as 48 tooth sample slices with approximately flat surfaces. Forty-two slices were used in the first section. The picosecond laser was 1,064 nm in wavelength, 3 W in power, and 10 kHz in repetition frequency. For a varying number (n=5–70) of focal plane additive pulse layers (14 groups, three repetitions each), two-dimensional scanning and ablation were performed on the dentin regions of the tooth sample slices, which were fixed on the focal plane. The ablation depth, d, was measured, and the quantitative function between n and d was established. Six slices were used in the second section. The function was used to calculate and set the timing of stepwise increments, and the single-step size along the focal plane normal was d micrometer after ablation of n layers (n=5–50; 10 groups, six repetitions each). Each sample underwent three-dimensional scanning and ablation to produce 2×2-mm square cavities. The difference, e, between the measured cavity depth and theoretical value was calculated, along with the difference, e_1, between the measured average ablation depth of a single-step along the focal plane normal and theoretical value. Values of n and d corresponding to the minimum values of e and e_1, respectively, were obtained. In two-dimensional ablation, d was largest (720.61 μm) when n=65 and smallest when n=5 (45.00 μm). Linear regression yielded the quantitative relationship: $d=10.547\times n-7.5465$ (R^2=0.9796). During three-dimensional ablation, e_1 was the smallest (0.02 μm) when n =5 and d=45.00 μm. The depth error was 1.91 μm when 450.00-μm depth cavities were produced. When ablating dentin with a three-axis picosecond laser scan-ablation device (450 μm, 3 W, 10 kHz), the number of focal plane additive pulse layers and step size along the focal plane normal was positively correlated with the single-layer and total ablation depth errors. By adjusting the timing of stepwise increments along the focal plane normal and single-step size when ablating dentin by using the numerically controlled picosecond laser, the single-step ablation depth error could be controlled at the micrometer level.

Keywords Picosecond laser · Numerically controlled ablation · Dentin · Ablation depth · Error control

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Introduction

Lasers have been used in stomatology since 1964. Thus far, laser technology has been used for treating oral mucosa and periodontal diseases, handling dentin hypersensitivity, controlling caries and removing carious tissues, preparing and disinfecting cavities, and preparing for tooth repair. Commercial dental lasers have obvious advantages in treating soft tissue diseases. However, they still have certain limitations in treating oral hard tissues, especially dental hard tissues. These limitations include insufficient ablation depth, easy production of micro-cracks, and difficulty in accurately controlling the three-dimensional appearance of the ablation surface [1, 2].

With appropriate parameters, ultra-short pulse lasers, including femtosecond and picosecond lasers [3, 4], could overcome the many clinical limitations of current commercial oral lasers. Their main reported [5] advantages include (1) high ablation efficiency, (2) significantly reduced mechanical or thermal damage, (3) relatively high accuracy in ablation depth control, (4) low noise levels, (5) reduced pain, (6) ability to change surface texture by controlling the shape and grating of the beam, (7) accurate control and safety (multi-photon treatment could ensure that the tissue underneath and around the focused light spot will not to be cut by the laser), and (8) potential to effectively cut all types of biological tissues as a result of the non-linear mechanism of ultra-short pulses.

Current studies in China and other countries mainly focus on the micro-morphology, thermal effects, and ablation efficiency of picosecond lasers on dental hard tissues [6–13]. However, there have been relatively few studies on the accurate control of ablation depth.

Materials and methods

Equipment and materials

Equipment and software

Laser system: picosecond laser (wavelength 1,064 nm; pulse width 15 ps; repetition frequency 10 kHz; power 3 W; size of the laser focal light spot 33.8 μm).

Numerically controlled laser galvanometer scanning system: a three-axis laser scanning device with mini numerical control galvanometer was developed and constructed by the authors (focal length of the lens 150 mm; light spot scanning speed of the two-dimensional scanning galvanometer 169 mm/s; light spot overlapping rate within and between scanning lines 50 %; minimum step size along *z*-axis 0.1 um; maximum step size 10 mm). The numerical control software was programmed by the authors.

Sample stage position control device: spiral micrometer (Mitutoyo Corporation, Kanagawa, Japan; accuracy: 0.01 mm).

Three-dimensional measuring device: three-dimensional laser scanning microscope (VK-X200, Keyence Corporation, Osaka, Japan).

Materials

Twenty-four freshly removed mandibular first molars were collected from the Oral and Maxillofacial Surgery Clinic, Stomatology Hospital of Peking University.

Methods

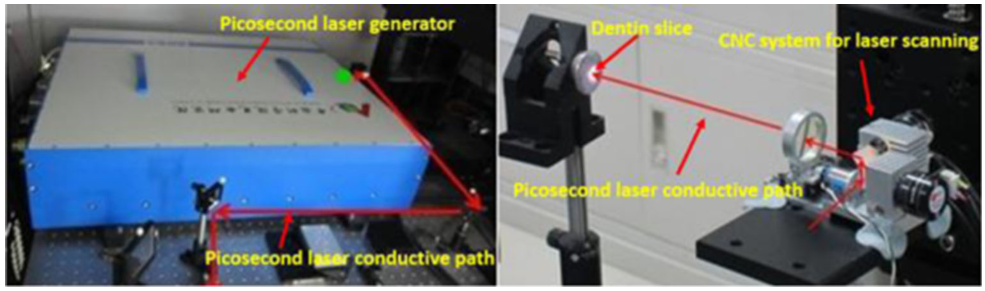
Sample preparation

Calculus and soft tissues on the dental surfaces were removed with an ultrasonic scaler, and the surfaces were rinsed clean with saline. The crowns and the roots were transversely cut along the cemento–enamel junction by using a diamond wire cutter (STX-202, Shenyang Kejing Instrument Co., Ltd., Shenyang, China). The diamond wire was moved parallel to the cross-section towards the crown, in order to produce 48 dental hard tissue slices 2 mm thick. The removal of enamel was monitored by the naked eye. The surfaces for ablation were approximately flat and perpendicular to the long axis of the crown. The two cross-sections were polished by hand, using 800-grit and then 1,000-grit waterproof sandpapers. The slides were submerged in formalin solution until they were used in the experiments. The samples were divided into two groups: 42 slices were used in two-dimensional scanning and ablation with a fixed focal plane position, six slices were used in three-dimensional scanning and ablation stepwise along the normal of the focal plane (hereinafter referred to as focal plane normal).

Two-dimensional scanning and ablation with fixed focal plane position

One sample was secured on the sample stage, which was fixed on a moving platform with spiral micrometers (two spiral micrometers were used to control the two-dimensional motions of the control platform). These ensured that the picosecond laser beam was perpendicular to the sample surface for ablation. By adjusting the platform using the spiral micrometers, the dentin surface for ablation was placed at the focal point in front of the center of the lens. The picosecond laser path was designed by software, as shown in Figs. 1 and 2. The frame was a square with an outer frame of 5 mm in width. The line spacing was 16.9 μm (with 50 % light spot overlapping rate). The inner frame was 3 mm in width. The control software calculated the number of frames to be scanned and controlled the two-dimensional scanning and ablation of the laser spot from the outer to inner frame with a fixed focal plane position. This motion path of the picosecond laser was applied to the dentin slice surface for *n* layers of additive pulse

Fig. 1 Experimental platform



(one layer of additive pulse means the completion of one ablation motion from the outer to the inner frame by the laser; n represents the layer number). The remaining 13 dentin slice surfaces were scanned for n=10–70 layers (in steps of 5).

Measurement of two-dimensional scanning and ablation depth

A three-dimensional laser shape measurement microscope was used to measure the actual ablation depth of each dentin slice, d (the actual two-dimensional ablation depth with fixed focal plane position). Forty points were selected on the bottom surface of each dentin slice (the points were distributed as evenly as possible on the cut surface). With the upper surface as the reference, the depths of the 40 points were measured to calculate an average. Repeated cuts were made in three slices for each n value to calculate an average. Hence, the actual ablation depth corresponding to each n value was obtained. The maximum d value in two-dimensional ablation was obtained, and the curve that quantitatively described the relationship between d and n was drawn.

Three-dimensional scanning and ablation with stepwise increments along focal plane normal

Parameters such as the focal plane two-dimensional scanning path were set as described in the "Two-dimensional scanning and ablation with fixed focal plane position" section. The timing for stepwise increments along the normal was set for n at 5 to 50 layers (in steps of 5). The single-step size was set according to the relation d_0=10.547×n–7.5465, which was obtained from linear regression analysis of the quantitative relationship between d and n. The number of steps, t, was set to 9, 4, 3, 2, 2, 1, 1, 1, 1, and 1. Ten square cavities (2× 2 mm) were cut into each dentin slice. This was repeated in six sample slices.

Measurement of three-dimensional ablation depth

A three-dimensional laser shape measurement microscope was used to measure the actual total ablation depth, d_2, of each square cavity by using the method described in the "Measurement of two-dimensional scanning and ablation

Fig. 2 Two-dimensional motion paths of picosecond laser spot

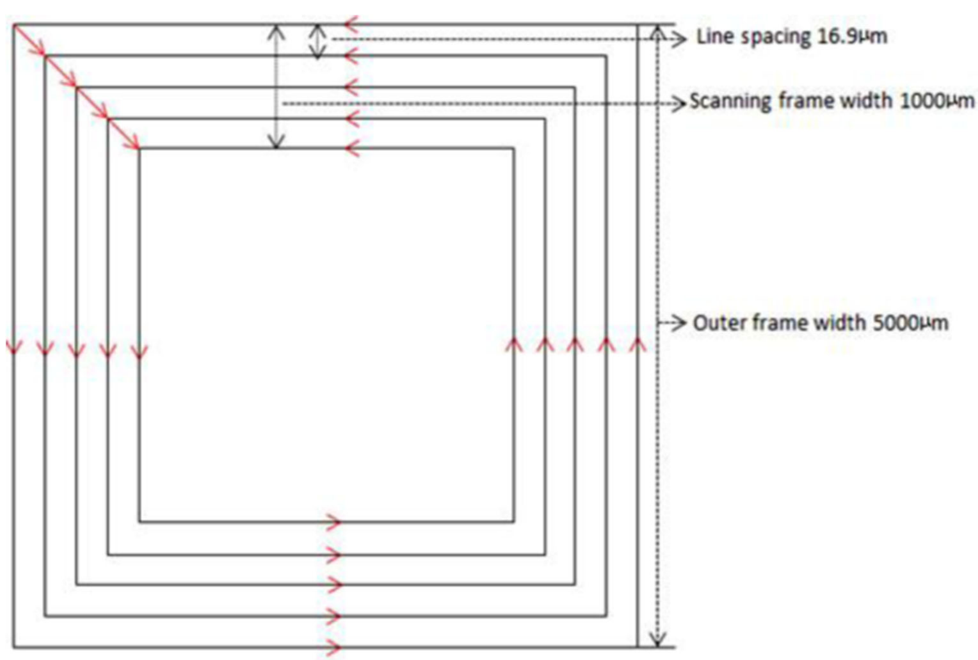


Table 1 Correlation between the number of two-dimensional scanning layers and the ablation depth in dentin

n (layer)	*d* (μm)
5	55.14
10	114.57
15	164.29
20	221.35
25	241.63
30	301.00
35	348.97
40	394.40
45	420.57
50	505.67
55	532.45
60	680.23
65	720.61
70	713.64

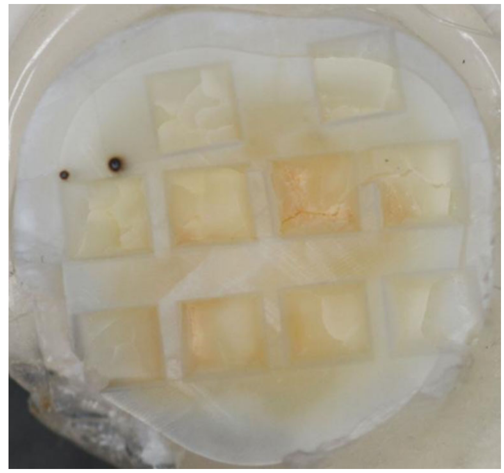


Fig. 4 Square cavities formed by ablating dentin three-dimensionally, using picosecond laser

depth" section. The theoretical total ablation depth is $d_1 = d_0 \times (t+1)$, where d_0 is the theoretical single-step ablation depth. The error in total depth was e, and the error in focal plane single-step ablation depth was $e_1 = (d_2/t+1) - d_0$. The relationships among e, e_1, n, and d were analyzed.

Results

Two-dimensional ablation

Table 1 contains the values of n and d for each laser ablation in dentin. The maximum d was 720.61 μm when n=65.

Within the range of the maximum ablation depth, d increased as n increased. Linear regression analysis revealed that the function equation and correlation coefficient were $d = 10.547 \times n - 7.5465$ and $R^2 = 0.9796$, respectively. A graphic representation of the quantitative relationship between n and d is shown in Fig. 3.

Three-dimensional ablation

Ten square cavities ablated three-dimensionally in dentin by a picosecond laser are shown in Fig. 4. These correspond to the timings of ten steps and single-step sizes.

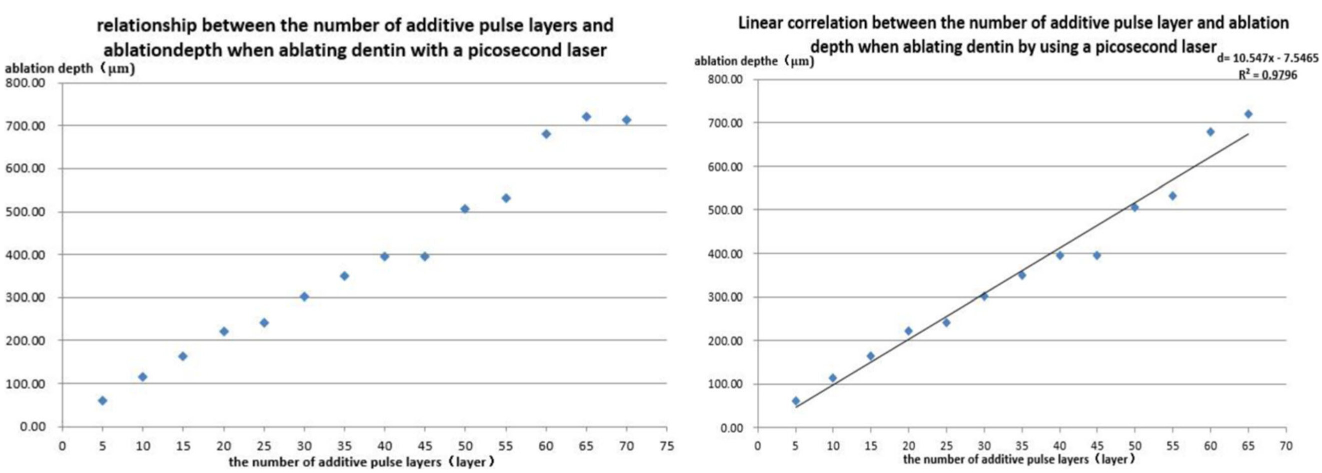


Fig. 3 Graphic representation of the quantitative relationship between the number of focal plane additive pulse layers and ablation depth during two-dimensional scanning and ablation of dentin using picosecond laser. *Left*, relationship between the number of additive pulse layers and ablation depth when ablating dentin by using a picosecond laser; *y*-axis; *right*, linear correlation between the number of additive pulse layer and ablation depth when ablating dentin by using a picosecond laser. For both panels, the *y*-axis shows ablation depth (μm), and the *x*-axis denotes number of pulse layers (*layer*)

Table 2 Parameter settings and results of three-dimensional dentin ablation

n (layer)	d (μm)	t (time)	d_1 (μm)	d_2 (μm)	e_1 (μm)	e (μm)
5	45.00	10	450.00	451.91	0.02	1.91
10	98.00	5	490.00	540.72	1.01	50.72
15	151.00	4	604.00	669.43	1.09	65.43
20	203.00	3	609.00	744.67	2.26	135.67
25	256.00	3	768.00	946.42	2.38	178.42
30	309.00	2	618.00	858.52	4.01	240.52
35	362.00	2	724.00	1024.48	4.30	300.48
40	414.00	2	828.00	1189.86	4.52	361.86
45	467.00	2	934.00	1374.24	4.88	440.24
50	520.00	2	1040.00	1532.76	5.03	492.76

When n=5, e_1 was the smallest at 0.02 μm

A three-dimensional laser shape measurement microscope was used to measure d_2. The calculated values of d_1, e, d_0, and e_1 are shown in Table 2.

Conclusion

The experimental results showed that when ablating dentin two-dimensionally by a numerically controlled picosecond laser, the number of focal plane additive pulse layers and ablation depth had an overall positive correlation. In three-dimensional ablation, when the single-step size along the focal plane normal was set to be the same as the two-dimensional scanning and ablation depth, the number of picosecond laser focal plane additive pulse layers and single-step size along the normal were positively correlated with the depth errors for single-step ablation and total ablation.

By adjusting the timing of stepwise increments along the focal plane normal and the single-step size, the depth error of ablating human dentin by a numerically controlled picosecond laser could be controlled within 0.02 μm per step. When the timing of stepwise increments and single-step size along the focal plane normal were set at five additive pulse layers and 45.00 μm, respectively, square cavities with a depth of 450.00 μm were produced while the error of dentin ablation depth could be controlled to 1.91 μm.

Discussion

Academics from China and other countries have been studying the application of lasers in oral medicine since the 1960s and have tried to replace traditional scalpels and dental hand pieces. Previous studies have found that for both thermal and Waterlase ablation, long-pulse lasers (millisecond, microsecond, or nanosecond lasers) have shortcomings that are very difficult to overcome when ablating dental hard tissues. With relatively long pulses, the excess energy can spread to adjacent tissues. The resultant heat and shockwaves can also cause micro-cracks in the ablation surface, irritate the pulp nerve, and leave the patients with various degrees of pain. When using a Waterlase laser, the water molecules within the dental hard tissues could burst and cause the surrounding tooth tissues to disintegrate, leading to loss of control over ablation accuracy. This treatment, which does nothing to improve the smoothness of the ablation surface, could damage healthy dental tissues. The light ablation properties of picosecond and long-pulse lasers are different. The main manifestation is plasma-induced ablation, which is produced by the ionization of the plasma itself. Because this process is extremely fast, it passes on very little energy, and hence causes very little thermal damage to the surrounding tissues [13]. Smooth and clearly defined tissue separation surfaces could be seen by using plasma-induced ablation, without any trace of thermal or mechanical damage.

Although picosecond lasers have certain advantages compared with long-pulse lasers, their laser energies show non-uniform Gaussian intensity distributions. Previous studies and pilot experiments have revealed that the ablation surface may have the appearance of volcanic craters, which makes it difficult to accurately control the depth when ablating dental hard tissues. However, this difficult problem can be solved by controlling the number of two-dimensional scanning layers with fixed focal plane position and a single-step size along the focal plane normal. In this experiment, we introduced a numerical control device that could accurately control the laser to complete the entire scanning process along a previously planned motion path. By accurately controlling the timing of stepwise increments and the single-step size along the focal plane normal when ablating dentin by a laser, the ablation depth error was controlled within the level of micrometers. The relatively low ablation efficiency remains a difficult issue in the current experiments. Future research could include adjusting the laser parameters to increase ablation efficiency or planning new ablation paths to reduce the actual dental tissue volume that needs to be ablated during dental preparation.

Previous research has shown that picosecond lasers with the same parameters ablate dentin with higher efficiency than they do enamel. LiZerilla et al. [14, 15] showed that with the same energy density, the efficiency of ablating dentin by picosecond laser is about eight times that of ablating enamel. The main reason is that compared with enamel, dentin contains a large amount of collagen, which makes it easier to ablate. In this experiment, we excluded the enamel layers in the samples and only studied dentin, in order to avoid the impact of unevenness in the materials on the results. In future studies, we will continue to explore the features of picosecond laser-enamel interactions and complete the full-crown tooth preparation process of an intact ex vivo tooth by using a picosecond laser.

Different laser parameters such as surface scanning speed, repetition frequency, and output power could result in different effects on dental hard tissues as well as in carbonation and raised temperature (roughly from 5 to 100 °C). Surface scanning speed has important effects in laser ablation processes. Relatively lower scanning speed could lead to a higher overlapping rate and a greater likelihood of the carbonation of dental hard tissue surfaces and significantly elevated temperature. In contrast, high-speed scanning produces relatively little thermal damage. Fahey et al. [16] showed that when ablating dentin by using a picosecond laser at a scanning speed of 1 mm/s, the high overlapping rate caused dentin surface carbonation and significantly raised temperature.

Light spot diameter and scanning speed both affect light spot pulse overlapping rate. Based on the result in the pilot experiment, the light spot overlapping rate in this experiment was set as 50 %. Dentin ablation by laser gave a relatively good result without carbonation. Studies on increasing temperature would be further expanded in future experiments.

Picosecond laser beams follow a Gaussian distribution and may effectively ablate dentin on either side of the light direction (hence the focal plane normal), given that the energy density exceeds the threshold of effective dentin ablation. In this experiment, during two-dimensional scanning and ablation of dentin with fixed focal plane position, the ablation depth was maximal at 65 additive pulse layers and decreased at 70 additive pulse layers. We believe the main reason for this was that the energy density gradually decreased with increasing ablation depth until it was lower than the threshold for dentin ablation and could not perform effective ablation. However, because laser energy works on the ablation surface, this may cause melting and re-solidification of the surface hard tissue debris and of dentin itself. The accumulation on the ablation surface would stop the depth from increasing and cause it to decrease instead.

In the three-dimensional ablation of dentin to form square cavities, the number of steps along the focal plane normal was set to gradually decrease as the number of single additive pulse layers increased. The main reason for this was that when using a picosecond laser in two-dimensional dentin scanning, the volcanic crater-like appearance of the lower surface produced by ablation was more obvious the greater the number of additive pulse layers. If the number of steps along the focal plane normal in each three-dimensional ablation was set to 10, the resultant error in ablation depth measurement would be relatively large. It was also possible to cut through the 2mm sample. At the same time, three-dimensional ablation was not performed at n=55–70 primarily because when n was over 50, the ablated surfaces showed volcanic crater-like appearances that made it difficult to measure the depth accurately. There were mainly two reasons for setting the minimum additive pulse layer number to be five. One was that with the measurement method used in this study, the depth would be too small and the measurement error large if the layer number was smaller than five. The other was that ablation at 10 mm (normal height of a tooth crown) with this parameter yielded a theoretical cumulative error of 44 μm. Currently, the clinical accuracy of manual tooth preparation depth often exceeds the 100 μm level. Hence, tooth preparation with five single additive pulse layers would already satisfy the clinical requirement as well as the requirement by the FDA (USA) that the total crown edge adhesion level should be within 120 μm. However, we are still far from the optimal accuracy of edge adhesion level (30 μm). We will continue to refine the number of additive pulse layers in the next step and attempt to reach the optimal accuracy requirement in ablation depth.

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