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ORIGINAL ARTICLE

Heat generation caused by ablation of dental hard tissues with an ultrashort pulse laser (USPL) system

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Abstract Heat generation during the removal of dental hard tissues may lead to a temperature increase and cause painful sensations or damage dental tissues. The aim of this study was to assess heat generation in dental hard tissues following laser ablation using an ultrashort pulse laser (USPL) system. A total of 85 specimens of dental hard tissues were used, comprising 45 specimens of human dentine evaluating a thickness of 1, 2, and 3 mm (15 samples each) and 40 specimens of human enamel with a thickness of 1 and 2 mm (20 samples each). Ablation was performed with an Nd:YVO_4 laser at 1,064 nm, a pulse duration of 9 ps, and a repetition rate of 500 kHz with an average output power of 6 W. Specimens were irradiated for 0.8 s. Employing a scanner system, rectangular cavities of 1-mm edge length were generated. A temperature sensor was placed at the back of the specimens, recording the temperature during the ablation process. All measurements were made employing a heat-conductive paste without any additional cooling or spray. Heat generation during laser ablation depended on the dental hard tissue (enamel or dentine) and the thickness of the respective tissue ($p<0.05$). Highest temperature increase could be observed in the 1-mm thickness group for enamel. Evaluating the 1-mm group for dentine, a significantly lower temperature increase could be measured ($p<0.05$) with lowest values in the 3-mm group ($p<0.05$). A time delay for temperature increase during the ablation process depending on the material thickness was observed for both hard tissues ($p<0.05$). Employing the USPL system to remove dental hard tissues, heat generation has to be considered. Especially during laser ablation next to pulpal tissues, painful sensations and potential thermal injury of pulp tissue might occur.

Keywords Heat generation · Ultrashort pulse laser system · Dental hard tissues · Enamel · Dentine · Temperature measurement

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Introduction

Laser technique is used in the oral cavity for a variety of dental treatment procedures such as surgical excisions or incisions of soft tissues, gingivectomy, frenectomy, and a variety of soft tissue periodontal procedures [1]. Application of laser irradiation to dental hard tissues has been described for the treatment of dental caries [2] and as an alternative for formocresol in pulpotomy of deciduous teeth [3]. In general, laser applications are considered as alternative methods for a variety of dental treatment procedures as there is no mechanical vibration, causing less painful sensations [4].

Employing ultrashort pulse lasers (USPL), ablation of oral hard and soft tissues is reported with minimal collateral damage and high precision at sufficient ablation rates [5].

Comparing nanosecond and picosecond laser ablation in enamel, advantages toward the use of ultrashort laser pulses in dentistry are suggested [6]. Even laser energy concentration in femtosecond pulses has been evaluated for enamel, dentine, and cementum ablation [7], and microcavities with very precise edges could be observed [8]. The present ultrashort pulse laser technology is based on systems in the range of 1 μm wavelength and pulse durations of femtosecond to picosecond [9]. The laser–tissue interaction is referred to as plasma-induced ablation or, in a more general way, photodisruption [10, 11]. Typical energy settings are in the range of up to 100 μJ, 500 kHz at a focus diameter of about 30 μm coupled with a scanner system.

Unpublished preliminary studies indicate that the USPL technology might be an effective tool for the ablation of oral hard and soft tissues with accurate preparation outlines and the lack of thermal damages around the prepared cavities. However, a temperature increase during the removal of dental restorative materials could be observed [12], which was dependent on both the kind and the thickness of the restorative material under study. Thus, heat generation has to be considered during ultrashort pulse laser ablation. Further studies have to evaluate the clinical impact of this in vitro measured temperature increase with respect to dental tissues, the pulp microcirculation, and intra-pulp chamber temperature changes.

Heat generation during a tooth preparation procedure may potentially have a hazardous impact on the dental pulp and damage the tissue irreparably if uncontrolled [13, 14]. In vivo examination of dental pulps exposed to heat revealed that an elevation in temperature to 39–42 °C resulted in an increase in circulation (hyperemia) [15]. If temperatures of 46 to 50 °C were maintained for several seconds, thrombosis and a standstill of circulation occurred. An in vivo study correlated the rise in temperature on the surface of the tooth with an internal rise in temperature [14]. It could be demonstrated that an increase in pulpal temperature to 42.2 °C led to pulpal necrosis, which could be observed in some cases, and a further rise in temperature caused necrosis in more than half of the teeth in their animal model.

As a consequence, it is essential to assess the possible heat generation capacity for any new tooth preparation technology to avoid unwanted side effects. Among the most studied laser systems, namely the ruby, CO_2, and neodymium-doped yttrium aluminum garnet (Nd:YAG) lasers, heat generation above the pulpal tolerance is described when used on mineralized tissues [16, 17]. Employing an erbium-doped yttrium aluminum garnet (Er:YAG) laser and a conventional high-speed handpiece, similar temperature increases could be observed for preparation procedures under water cooling [13]. Evaluating a 10-ps neodymium-doped yttrium vanadate (Nd:YVO_4) laser for cavity preparation in enamel and dentine, a temperature increase of up to 30 K for enamel and 39 K for dentine could be measured depending on the choice of laser and scanning parameters [18].

The biological aspect of heat generation is one of the most important issues to be evaluated before introducing a novel technology into daily treatment concepts. Preliminary observations make the USPL technology a promising tool for a variety of dental applications such as tooth cavity preparation, caries therapy, removal of restoration materials, periodontal treatment, or also endodontic and implant therapy. Unfortunately, there is only low evidence for the use of USPL systems in dentistry, as systematically performed and statistically analyzed studies are rare. Thus, the aim of the study was to perform an intraexperimental comparison of heat generation in dental hard tissues following laser ablation using an ultrashort pulse laser system, testing the hypothesis of thermal changes being dependent on both the kind and the thickness of the hard tissues under study.

Materials and methods

Forty-five freshly extracted human teeth (molars and incisors) were collected from different patients and stored in a physiological saline solution. The study was conducted in full accordance with the declared ethical principles (World Medical Association Declaration of Helsinki, version VI, 2002). Patients were informed that the extracted teeth would be used in an in vitro study. Employing these teeth, a total of 85 specimens of dental hard tissues were prepared, comprising 45 specimens of human dentine evaluating a thickness of 1, 2, and 3 mm (15 samples each) and 40 specimens of human enamel with a thickness of 1 and 2 mm (20 samples each). Hard tissue specimens were reduced to the respective thickness employing a water-cooled diamond saw (Exakt 300 CP Band System, Exakt Apparatebau, Norderstedt, Germany) with an accuracy of ±0.1 mm.

Laser irradiation

Laser irradiation was performed with an Nd:YVO_4 laser at 1,064 nm (modified Super Rapid, Lumera Laser, Kaiserslautern, Germany), a pulse duration of 9 ps, and a repetition rate of 500 kHz. The output power of the laser device was set to 6 W for all measurements. Laser irradiation time was 0.8 s with a delivered total energy of 4.8 J. Employing a scanner system with a scanning speed of 2,000 mm/s (Scan Cube 7, Scanlab, Puchheim, Germany), rectangular cavities of 1-mm edge length were generated (Fig. 1). The scan pattern consisted of individual horizontal lines with a spot size of 30 μm and a spacing of 12.5 μm. At the end of each line, the laser is automatically switched off and moves to the next line. The laser pulses need to overlap to perform homogeneous ablation patterns. This overlap is determined by scanning speed, repetition rate, and pulse diameter along each line, whereas the vertical overlap of two successive lines is affected by pulse diameter and line distance. Parameters for the present study led to an overlap of 44 % in horizontal and 84 % in vertical direction. The final cavity was the result of 12 scans. Prior to temperature measurements, a preliminary survey was performed to assess the ablation of both enamel and dentine samples. Employing an optical surface measuring system (MicroSpy, Fries Research and Technology, Bergisch Gladbach, Germany), the ablation depth of 26 specimens of each hard tissue was evaluated. Additionally, all samples were photographed employing a light microscope (Wild M8, Leica Mikrosysteme, Wetzlar, Germany).

Temperature measurements

A temperature sensor with an outer diameter of 0.5 mm was placed at the back of the specimens, recording the temperature during the ablation process (TDA 3000, Jumbo, Fulda, Germany). For all measurements, the tip of the temperature sensor was aligned to the optical path of the laser beam using a translation stage (xyz-table LWRE3, SKF Linearsysteme, Schweinfurt, Germany) with a 0.02-mm precision of adjustment. Thereafter, the sample under study was aligned to the focal laser spot employing another translation stage (xyz-table VT-80, Micos, Eschbach, Germany) with a 0.001-mm precision of adjustment (Fig. 1). The temperature sensor was placed in contact behind the sample preserving the optical path of the laser beam. All measurements were performed employing a heat-conductive paste (P12, Wacker Silicones, Drawin Vertriebs GmbH, Ottobrunn/Riemerling, Germany) without any additional cooling or spray. Temperature measurements started 1 s before starting the ablation process to assess the baseline value. Further temperature values were measured at intervals of 1 s until the maximum temperature reading decreased. The primary outcome measures were the maximum temperature increase and the time delay measured for the maximum temperature after starting laser ablation. All measurements were performed in a room where a constant temperature of 21 °C was observed by an air conditioning system.

Statistical analysis

A power analysis was performed prior to the study. Therefore, the effect size was set to 0.8 according to Cohen [19]. For an alpha error of 0.05 and a power of 0.8, a sample size of at least 15 specimens in each group was calculated. For each enamel and dentine specimens, three temperature measurements were performed to minimize the impact of any individual structural conditions. Subsequently, the mean value of these three measurements was used for further statistical calculations so that one sample could be considered as one statistical unit. For statistical analysis, normal distribution of the values was assessed with the Shapiro–Wilk test. Since not all data were normally distributed, values between the groups were analyzed with a nonparametric test (Kruskal–Wallis) and Mann–Whitney pairwise comparisons. Differences were considered as statistically significant at $p<0.05$. Box plot diagrams show the median, first and third quartiles, and minimum and maximum values (whiskers). Values of more than 1.5 to 3 times the interquartile range are specified as outliers and marked as data points. Values more than three times the interquartile range are specified as far outliers and marked as asterisks.

Results

Ablation depth

Profilometry showed a median ablation depth of 29 μm (min, 9 μm; max, 87 μm; interquartile range, 16 μm) in enamel and 43 μm (min, 14 μm; max, 80 μm; interquartile range, 35 μm) in dentine. Ablation depths were statistically significantly different ($p<0.05$, Mann–Whitney).

Maximum temperature increase

Heat generation could be observed during ablation of both enamel and dentine specimens. In general, enamel samples showed a higher maximum temperature increase than dentine samples ($p<0.05$). However, temperature increase depended on the thickness of the respective material ($p<0.05$) (Figs. 2 and 3). Lowest delta temperature values could be observed in the 3-mm thickness group for dentine (median delta value, 0.7 K; min, 0.5 K; max, 1.2 K) and in the 2-mm thickness group for enamel (median delta value, 7.3 K; min, 4.9 K; max, 23.9 K). Highest delta temperature values could be observed in the 1-mm thickness group for both dentine (median delta value, 6.8 K; min, 3.6 K; max, 13.6 K) and enamel (median delta value, 44.3 K; min, 29.3 K; max, 67.0 K).

Temperature increase delay

A time delay for the maximum temperature after starting laser ablation could be observed in all groups (Fig. 4). Highest latencies occurred in the 3-mm dentine group (median latency, 4.3 s; min, 2.2 s; max, 12.2 s; interquartile range, 3.3 s). A decrease of material thickness resulted in lower latencies of the maximum temperature in the dentine group (median latency dentine 1 mm, 2.0 s; min, 2.0 s; max, 3.8 s; interquartile range, 0.5 s; $p<0.05$). No difference could be observed for the 1-mm (median latency, 2.1 s;

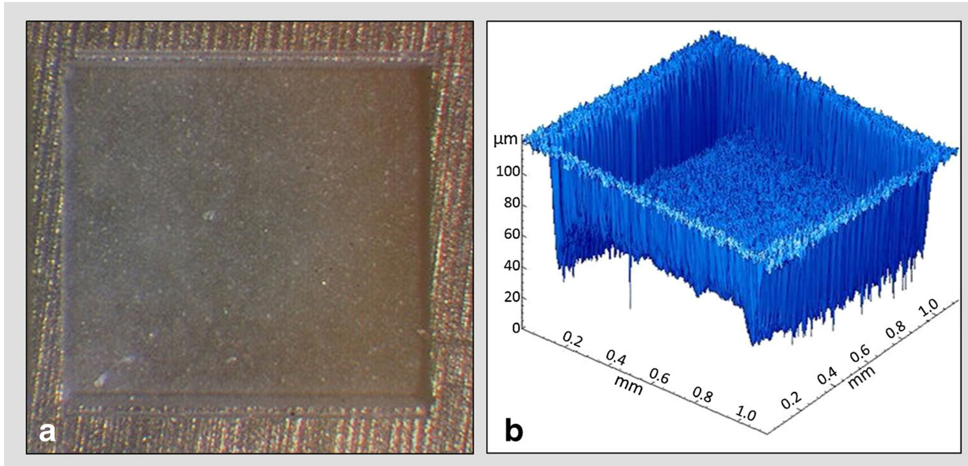

Fig. 1 Representative dentine specimen after laser ablation of a rectangular cavity with 1-mm edge length assessed with a light microscope (**a**) and a profilometer (**b**)

min, 1.8 s; max, 3.5 s; interquartile range, 0.2 s) and 2-mm (median latency, 2.2 s; min, 1.8 s; max, 6.0 s; interquartile range, 0.3 s) enamel specimens ($p>0.05$) with also no statistically significant difference to the 1-mm dentine group ($p>0.05$).

Discussion

The present study showed that heat generation during ablation of dental hard tissues with an ultrashort pulse laser system depends on the type and the thickness of the material. The amount of heat generation was significantly different in enamel and dentine with highest values for thin specimens. In general, enamel samples showed a statistically significant higher temperature increase than dentine specimens. In a previously published study, it could be demonstrated that heat generation occurred also during removal of restorative dental materials employing a USPL system [12]. These differences in heat generation with respect to diverse tissues and materials may be caused by a threshold nature of the ablation process, which is described for nickel employing picosecond laser ablation [20]. Moreover, a threshold caused higher temperature increase in the enamel group would be in accordance with the results of the ablation depth measurements. It could be shown that the ablation depth in enamel was statistically significantly lower than in dentine.

Employing a 10-ps $Nd:YVO_4$ laser, it could be shown that cavity preparation in enamel and dentine resulted in similar ablation rates and temperature increases in 1-mm-thick specimens [18]. At a fluence of 7.6 W/cm^2, an ablation rate of approximately 2.0 µm/pulse in enamel and 2.8 µm/pulse in dentine were measured. In the present study,

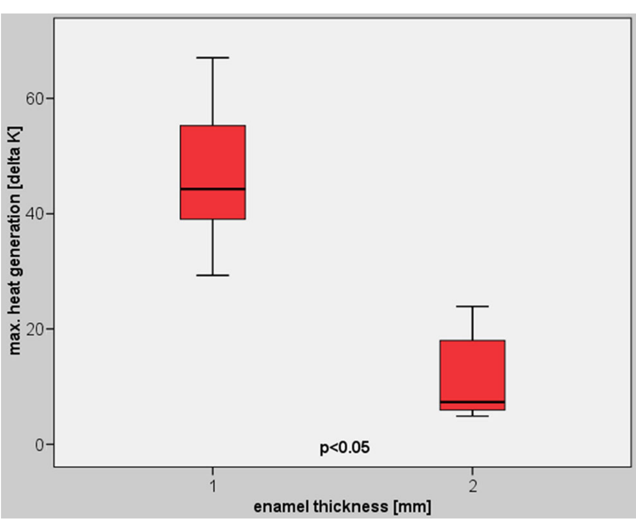

Fig. 2 Maximum heat generation during ablation of enamel specimens ($n=20$ per group). Temperature increase was higher for thin samples ($p<0.05$) with a maximum increase of 67 K in the 1-mm group

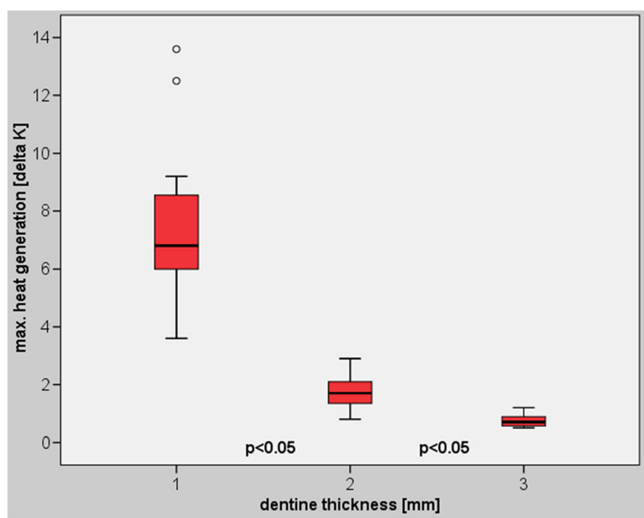

Fig. 3 Maximum heat generation during ablation of dentine specimens ($n=15$ per group). Temperature increase was higher for thin samples ($p<0.05$) with a maximum increase of 13.6 K in the 1-mm group

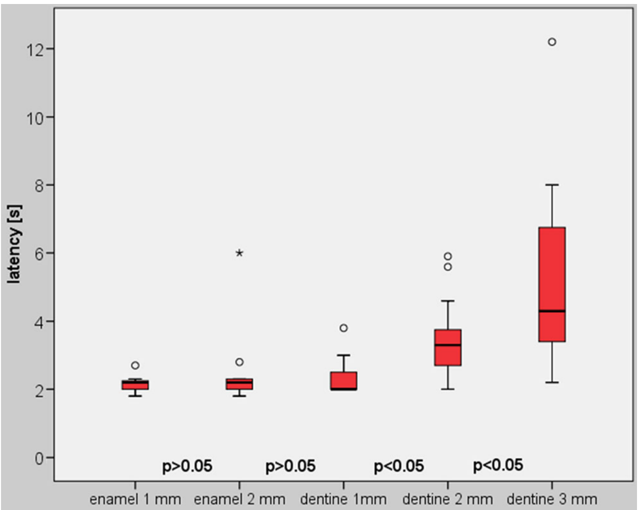

Fig. 4 Time delay measured for the maximum temperature after starting laser ablation of the different hard tissue specimens. Highest latencies occurred in the dentine group ($p<0.05$). An increase of material thickness resulted in higher latencies of the maximum temperature in the dentine group ($p<0.05$) with no difference in the enamel group ($p>0.05$)

a fluence of only 1.58 J/cm^2 was applied at the target surface and showed deeper cavities in dentine than in enamel. In hard tissue specimens with a thickness of 1 mm, an increase of 44.3 K in enamel and 6.8 K in dentine could be observed. A temperature increase of approximately 12 K for enamel and 6 K for dentine is reported for the 10-ps Nd:YVO$_4$ laser [18]. In this context, it is remarkable that the values observed in dentine are almost the same in both studies, whereas the temperatures measured in enamel differ immensely. Enamel providing the higher temperature increase than dentine is in agreement to the findings of the present study. However, using higher output powers than 2.9 W, the authors measured an increase of up to 30 K in enamel and 39 K in dentine [18]. The differences between these findings may have several reasons. In general, laser–tissue interaction depends on the applied fluence. Assuming that the observed temperature increases are a cumulative effect, it is important to note that the irradiation time in the present study was slightly higher. Employing two more repetitions of a single scanning pattern in the present study, a temporal difference of approximately 0.2 s or 25 % occurred. Moreover, both ablation rate and temperature increase also strongly depend on the scan parameters as well as on the water content of the samples. As the authors stated, a small water film at the sample surface drastically reduces the ablation rate since it absorbs the greater part of the incident laser energy [18]. Therefore, summarizing the results of both studies, a clinically significant temperature increase and the risk of a potential damage to the dental pulp cannot be excluded. Furthermore, the used irradiation times were only about 0.6–0.8 s and therefore much shorter than in clinical practice. Thus, even if a properly optimized additional cooling method might hinder the ablation speed, it should be considered for safety reasons.

Evaluating morphological and thermal aspects with respect to Nd:YAG picosecond laser ablation both in dentine and enamel, it could be shown that this technology is a safe tool for ablation of primary teeth in a broad range of operational parameters where temperature changes do not exceed 5.5 K [21, 22]. Employing a high-speed scanning ablation procedure to remove dental hard tissues, it could be demonstrated that enamel and dentine surfaces can be rapidly ablated by CO_2 lasers with minimal peripheral, thermal, and mechanical damage and without excessive heat accumulation [23].

For comparability reasons, in vitro laser ablation performed in the present study represents an exact temporally defined and reproducible stimulus. In vivo treatment times depend on the amount of the dental hard tissues to be removed. Thus, different laser activation times have to be considered, resulting in different amounts of total laser energy emitted. The present study did not assess temperature changes during the removal of hard tissues with a given thickness. These data have to be evaluated in further studies to assess a possible cumulative temperature effect exceeding the pulpal threshold of heat tolerance. Assessing temperature changes in the pulp chamber during halogen lamp exposure for a tooth whitening procedure, such a cumulative effect could be observed [24]. However, longer laser activation periods do not necessarily lead to a continuous temperature increase. For the removal of dental hard tissues with Er:YAG and Er,Cr:YSGG lasers, it could be shown that both lasers generated greater pulpal heat increase than diamond burs, but the temperature increase did not exceed more than 5.5 K in all groups [25]. As no data are available for the removal of dental hard tissues with a USPL system, further studies have to evaluate the effect of long-lasting treatment procedures employing the novel laser device.

Several studies assessed the impact of temperature increase on dental pulp tissue. Heat exposure resulted in hyperemia in the rat incisor pulp by thermal irritation of 39–42 °C [15]. Temperatures of 44 °C and above caused red blood cell aggregation. Maintaining temperatures of 46 to 50 °C for 30 s caused thrombosis and a standstill of circulation. An in vivo study correlated the rise in temperature on the surface of the tooth with an internal rise in temperature [14]. The authors found that an increase in pulpal temperature to 42.2 °C caused pulpal necrosis in 15 % and a rise in temperature to 47.7 °C resulted in necrosis in 60 % of the teeth in their Macaca Rhesus monkey model. In general, a temperature increase higher than 5.5 °C was concluded to cause an irreversible damage of the pulp tissue.

In the present study, a temperature increase of up to 67 K could be observed in the 1-mm enamel group. However, this must not necessarily be the amount of heat energy that reaches the pulp tissue. It could be shown that dentine hard tissue is a good thermal insulator that significantly reduced temperature rises associated with resin composite photocuring [26]. Additionally, another clinically important aspect has to be taken into account, as the pulp microcirculation might affect the temperature increase. Simulating an artificial microcirculation with a continuous water flow inside the pulp chamber, the impact of various light curing units was assessed [27]. When the simulated pulp microcirculation was absent, the temperature increases produced by all curing units except the conventional halogen lamp were large enough to be potentially harmful to the pulp. On the contrary, with the cooling effect of water flow inside the pulp chamber, all units proved to be safe for use. Thus, the importance of the cooling effect of simulated pulp microcirculation in the thermal behavior of the dentine was established [28]. Temperature increase in a specimen depends on the tooth–laser interaction. The process of USPL ablation divides into three distinct time periods: the laser, hydrodynamic, and thermal periods [29]. During the laser period, absorption occurs in an evanescent wave in the self-generated plasma at the tissue surface with very little hydrodynamic motion or heat conduction. The second time period covers the ablation hydrodynamics, resulting in a hard tissue removal. The thermal period describes the thermal energy left in the tooth which results in heat generation. In the present study, it could be shown that a temperature increase was dependent on the thickness of the specimens. Thus, variables like heat capacity and conductivity of dental hard tissues result in lower temperature increase in thicker enamel or dentine specimens. With respect to clinical relevance, no thermal damage should be expected when dental hard tissue in the outer dentine layer is ablated. However, the thermal component of USPL ablation has to be taken into account during an ablation procedure in deep dentine cavities, leaving only a thin dentine layer covering dental soft tissues.

Evaluating pulp chamber temperature increase during composite light activation with respect to dentine thickness, a study could show a significant temperature increase with the reduction in dentine thickness evaluating dentine disks of 0.5 to 2 mm [30]. This observation is similar to the findings of the present study assessing both enamel and dentine specimens. It can be explained by the thicker material samples that might lead to a different thermal diffusivity with respect to the individual thermal properties. Measuring the thermal conductivity of dental hard tissues, values show a higher conductivity for enamel [31] than for dentine [32]. This finding is in accordance with the results in the present study, as higher latencies for the maximum temperature after starting laser ablation could be observed in the dentine group. However, the present study did not aim to calculate reliable values for the heat conductivity of enamel and dentine, especially as not the whole energy input was converted to heat. Consequently, the thermal conductivity should be assessed in further studies which do not comprise the ablation of dental hard tissues.

The present study aimed to perform an intraexperimental comparison of heat generation during the ablation process of dental hard tissues. The hypothesis of temperature increase being dependent on both the kind and the thickness of the hard tissue could be proven. Thus, heat generation has to be considered during laser ablation. Painful sensations might occur especially during laser ablation next to pulpal tissues. As heat may cause irreversible damage to the dentine pulp complex, further studies have to evaluate the clinical impact of pulp microcirculation and the influence on intra-pulp chamber temperature changes. Possibly, a cooling procedure has to be used for ablation of hard tissues next to the dental pulp.

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