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# Weakening of dentin from cracks resulting from laser irradiation

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ABSTRACT

Cracking of tooth structure is a frequent mechanism of clinical failure necessitating treatment. Some laser conditions, particularly those without sufficient water cooling, may cause surface cracking of dentin. Surface cracks may serve as initiation sites for the onset of catastrophic fracture under mechanical stress, resulting in failure of the dentin. In this study, the hypothesis that laser initiated cracks result in lower bending strength of dentin was tested.

Dentin beam specimens were prepared from human molar teeth, 1.1 mm × 1.1 mm × ~9 mm, and divided into groups C (control), W (wet), D (dry) of 12 beams each. In groups W and D, the middle of each beam on one surface (buccal) was irradiated with either a Er-YAG or Q-switched Er-YSGG laser and measured under a microscope, noting the dimensions in the irradiated area and immediately adjacent to irradiated area. Each beam was placed in a mechanical testing machine in a four-point bend jig and tested with a monotonically increasing load at a displacement rate of 1 mm/min until failure. The bending strengths for groups C, W (Er-YAG laser) and D (Q-switched Er-YSGG laser) were, respectively, 141.6, 114.0, and 90.9 MPa. A one-way ANOVA determined a significant difference between groups C and D, $p < 0.001$. *Conclusion*: The Q-switched Er-YSGG laser without water caused cracks in the surface that significantly decreased the bending strength of dentin.



## 1. Introduction

Cracking of teeth is a common occurrence resulting in damage which often necessitates extensive treatment [1]. Restorative treatment can predispose teeth to fracture by cutting cavity preparations which weaken teeth, particularly when multiple surfaces are involved [2]. The shape of the cavity may be a factor, for example, sharp corners can act as locations of stress concentration [3,4]. Cavity preparations can be optimized to reduce fracture [5].

The technique of cutting may be another factor, as some cutting instruments may result in more roughness and superficial cracking of the cavity wall. Recent advances in use of lasers to ablate tooth structure have resulted in increased clinical use of lasers for this purpose. Many types of lasers have been proposed for ablating hard tissues with varying wavelength, pulse duration and repetition rates.

Several lasers have been tested for this purpose and various laser parameters have been used. Some of the concerns with using lasers for ablation include thermal damage to the

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**Table 1 – Parameters associated with the laser irradiation and the resulting effect on bending strengths.**

| Group | Laser | Frequency (Hz) | Spot size (μm) | Fluence (J/cm$^2$) | Pulse duration (μs) | Distance between scans (μm) | Scanning distance between pulses (μm) | Bending strength in MPa (S.D.) | Statistical grouping |
|---|---|---|---|---|---|---|---|---|---|
| C | None | N/A | N/A | N/A | N/A | N/A | N/A | 141.6 (6.9) | A |
| W | Pulsed Er-YAG laser with water spray | 5.0 | 300 | 6 | 135 | 70 | 50 | 114.0 (24.5) | A and B |
| D | Q-switched Er-YAG laser without water spray | 3.0 | 300 | 5 | 0.5 | 70 | 83 | 90.9 (32.6) | B |

pulp and peripheral thermal and mechanical damage to tooth structure, particularly generation of cracks on the irradiated surfaces [6]. Suffice to say that any formation of cracks resulting from such laser treatments would severely weaken the tooth by lowering the stress to cause it to fracture.

Accordingly, the purpose of this study was to measure the differences in the fracture strength in bending strength of dentin beams irradiated under different laser conditions, specifically using a Er-YAG laser under wet conditions and a Er-YSGG laser under dry conditions. The hypothesis to be tested is that different laser conditions result in different levels of mechanical damage in dentin, which is reflected in changes in bending strength of dentin.

## 2. Materials and methods

### 2.1. *Specimen preparation*

Human molars were collected according to a protocol approved by the Institutional Review Board and sterilized by gamma radiation [7]. The teeth were sectioned in the bucco-lingual direction and then in the mesio-distal direction to prepare beams approximately 1.5 mm × 1.5 mm × 11 mm. These were shaped by wet abrasive paper to a final shape of 1.1 mm × 1.1 mm × ~9 mm, finishing with 600 grit. The majority of the teeth used only produced one beam per tooth, a few produced two beams. The orientation of the beams was marked on one edge at the end of the beam to denote the buccal surface. The samples, which were stored in water throughout the experiment, were divided into one control and two experimental groups. In the experimental groups, the middle of the buccal surface of each beam was irradiated by laser light using parameters summarized in Table 1.

### 2.2. *Tissue irradiation and laser parameters*

Samples were irradiated using a free-running Er-YAG laser and a Q-switched Er-YSGG laser. The lasers were originally manufactured by Schwartz Electro-optics (Orlando, FL). Er-YAG pulses of 35 μs pulse duration were generated using a Analog Modules (Longwood, FL) Model 8800 V variable pulse power supply that generates a square shaped flash lamp drive pulse. The second laser was a Er-YSGG (2.79 μm) laser system manufactured by Schwartz Electro-optics, (Orlando FL) operating in Q-switched mode (pulse duration 500 ns). The rotating-mirror Q-switch was custom manufactured (Shiva Laser, Los Angeles, CA). The laser energy was measured and calibrated using laser calorimeters (Model ED-200-Gentec, Quebec, Canada). The beam diameter at the position of irradiation was measured by scanning with a razor blade across the beam. The laser beam was Gaussian shaped, i.e., operating in a single TEM00 mode and was defined with a $1/e^2$, where e = 1/ln, or ~2.72, beam diameter. The laser pulse duration was measured with a room temperature HgCdTe detector, Boston Electronics Model PD-10.6-3 (Boston, MA) with a response time of a few ns.

The Gaussian shaped ($TEM_{00}$) laser spots had a 300-μm diameter and the laser was continuously scanned across the sample and laser pulse repetition rates of 5 and 3 Hz were

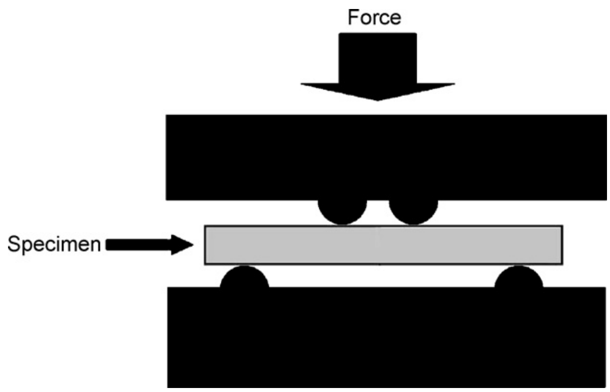


**Fig. 1 – Diagram of four-point bending apparatus.**

used (see Table 1). Eight parallel overlapping ablation scans were performed on each sample with a spacing of 70 μm between scans to produce an area of approximately 1 mm$^2$ using a computer-controlled motion control system ESP-300 (Newport, Irvine, CA) incorporating two 850G actuators.

The samples were kept well hydrated before irradiation and a low volume/low pressure air actuated fluid spray delivery system consisting of a EFD 780S spray valve, a Valvemate-7040 controller and a fluid reservoir from EFD Inc. (East Providence, RI) was used to provide a uniform spray of fine water droplets on the tissue surface during irradiation.

One group (W) was irradiated with short pulse (35 μs) Er-YAG laser at ablative conditions accompanied with water spray and the second group (D) was irradiated with a Q-switched Er-YSGG laser with a pulse duration of only 0.5 μs under non-ablative conditions known from previous studies to introduce cracks at the surface of the beam in dry conditions. The very short sub-microsecond Q-switched Er-YSGG laser pulses caused loud acoustic effects. It was anticipated that this would cause some mechanical damage such as micro-cracking.

The uniformity of the irradiation pattern was controlled by a two-dimensional scanning stage to which the beams were attached, as previously described [8]. The irradiated area included the entire width of the beam and approximately 1 mm of the length. A total of linear 8 laser ablation scans were performed, each scan spaced 70 μm apart to produce a square shaped laser irradiated surface.

### 2.3. *Mechanical testing*

After irradiation, the beams were examined in an optical microscope and the cross-section was measured in the irradiated zone, or the middle of the beam for the control group. Next, the beams were placed in a four-point bending apparatus and tested for breaking strength.

All testing was performed on a factory-calibrated ELF 3200 mechanical testing machine (EnduraTEC, Minnetonka, MN) in a custom-built four-point bend rig, made from Delrin (Fig. 1). The loading points were spaced 1.8 and 7.2 mm apart; the interface was centered between them. The spacing of the loading points was determined by the size of the beams, which are limited by the size of a human molar tooth. Each beam was positioned so that the irradiated area, which was visually discernible, was centered between the inner loading points. The irradiated area was placed facing down in the rig, and thus exposed to tensile stress during testing. Bending strengths, $\sigma_b$ (in MPa), were computed from the maximum load $P$ (in N), to cause failure, using the standard relationship (ASTM E855/1984):

$$\sigma_b = \frac{3PS}{bh^2} \times 10^6 \qquad (1)$$

where $S$ is the spacing (in meters) between upper and lower loading points, and $b$ and $h$ are, respectively, the specimen width and thickness (in meters). Mean bending strengths were calculated for each group of samples and tested for significant differences by a one-way ANOVA test.

## 3. Results

Although little or no ablation was noted for the dry Er-YSGG irradiated samples in group D, a small amount of ablation was observed on the wet Er-YAG irradiated surface in group W, decreasing the cross-section of the beam by approximately 0.08 mm. This area was approximately 0.5 mm wide, with the shape of a uniform shallow depression, rather than a notch. The 0.08 mm decrease in cross-section was measured by examining of the side view of the beam in a microscope. Examples of the irradiated areas are shown in Figs. 2 and 3. In group W, the irradiated area was clearly discernible visually, but no cracks were noted on the top or side surfaces (Fig. 2a and b). In group D, many samples had clearly visible cracks on the top surface, sometimes extending along the side surface (Fig. 3a and c). The crack pattern was irregular and not consistent from sample to sample, but in the side view, several cracks were observed in the range of 131–330 μm. However, no notching or depression was noted in the side view of samples in group D.

The mean bending strengths for each group are shown in Table 1. The control group (C) had the highest mean bending strength at 141.6 MPa. After wet Er-YAG irradiation (group W), the bending strength was reduced by almost 20% to 114.0 MPa; after dry Er-YSGG irradiation (group D), this reduction in strength was even larger, specifically by more than 35% to 90.9 MPa. There was a significant difference between groups C and D. The failure pattern was also differed somewhat among the three groups. In group C, the beams broke into 2–5 pieces (mean 2.92); in group W into 2–4 pieces (mean 2.67) and 2–4 pieces in group D (mean 2.25).

The location of the multiple failures appeared to be random in the control group C, but one of the cracks was generally near the middle of the beam. In group W, one of the cracks was usually in the irradiated area as shown in Fig. 2c. In group D the failure location corresponded to the cracks observed after irradiation (Fig. 3b and d).

## 4. Discussion

The ideal instrument for removing caries and preparing cavities should be selective for caries and produce a preparation

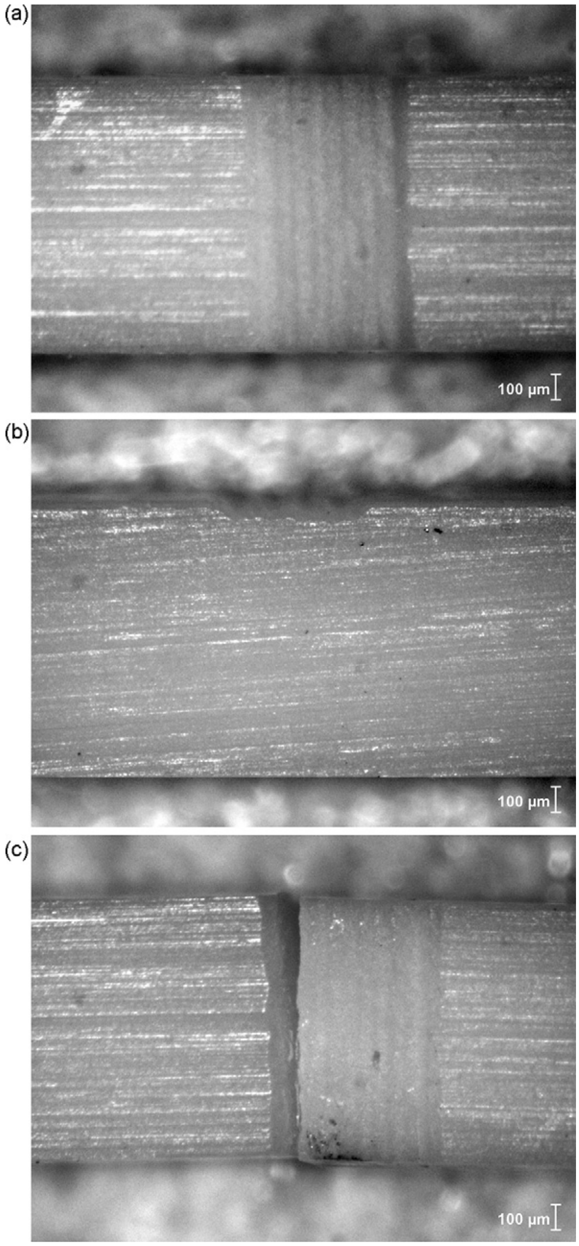


**Fig. 2 – (a) Irradiated surface, group; (b) side view, group W, showing ablated area at top; (c) same sample from group W after breaking, with failure location at the edge of the irradiated area.**

which does not adversely affect the tooth structure or the pulp and a surface which is receptive to bonding of restorative materials. Lasers have been studied in recent years as possible replacement for rotary cutting instruments in dentistry. However, there are concerns with efficiency and possible thermal damage to dentin or the pulp, as well as the bonding characteristics of the resulting surfaces.

The present study confirms our earlier findings that certain laser conditions cause thermal and mechanical damage to tooth structure during irradiation, while other conditions cause little or no peripheral damage [6]. This damage is not trivial, resulting in 20–35% reductions in the bending strength of the dentin. The variables which affect the level of peripheral damage in irradiated tooth surfaces include laser wavelength, pulse duration, fluence and water application rate. Two parameters were varied in the present study, specifically pulse duration and the presence of water.

Cracks are often observed during the irradiation of dental hard tissues. Short laser pulses generate strong elastic waves in the solid being ablated due to the transient thermal shock created by the laser heating, thermal expansion, and the recoil of the ablation products [9,10]. Such stress waves propagate through hard tissue possibly creating cracks and fractures. Tensile stresses are generated by the reflection of compressive thermoelastic stresses at the free surface of the dentin specimens. The magnitude of those stresses depends on the incident laser fluence, spot size and the laser pulse duration. The highest stresses are generated for laser pulses that satisfy the stress confinement condition where the laser pulse duration is less than the rate of stress wave propagation through the heated area. This is the product of the longitudinal speed of sound in dentin and the absorption coefficient [11]. For dentin the longitudinal sound velocity ranges from 2800 to 4000 $cm^{-1}$ [12] while the absorption depth at 2.79 μm is on the order of 10 μm [13], therefore stress confinement is in the tens of nanosecond range. The Q-switched Er-YSGG laser generates laser pulses of 500 ns duration which ten times longer than stress confinement. The laser pulses used in this study that does not produce cracks the Er-YAG laser with a 35-μs pulses duration has laser pulses a factor of 1000 times longer. We have directly observed crack formation and propagation in dental enamel in real time using near-IR imaging [14] by local dehydration and overheating the enamel by delivering laser pulses at a high pulse rate.

Past studies of shown that Q-switched Er-YSGG laser pulses produce a large number of small micro-cracks peripheral to the ablation site when a water spray was not used probably due to thermoelastic stresses generated by a combination of the mechanisms mentioned above [6]. Therefore we chose this laser irradiation condition to deliberately maximize mechanical damage and crack formation. Longer Er-YAG laser pulses of 35 μs were chosen with a water spray as the laser condition that minimizes thermal and mechanical damage to dentin. This second set of "ideal" laser parameters has yielded bond strengths (shear) on dentin surfaces equivalent to the conventional 35% phosphoric acid etch without post-ablation acid etching [15].

Mechanically, the presence of these laser-induced cracks can markedly reduce the fracture resistance of the tooth. Quantitatively, this can be estimated using simple fracture mechanics considerations which relate the fracture toughness, $K_c$, to the stress to fracture, $\sigma_f$, through expressions of the form $K_c \sim Q\sigma_f(\pi a)^{1/2}$, where $a$ represents the size of the largest defect (crack) present, and Q is a geometrical constant of order unity. For dentin, $K_c \sim 1.8\,\text{MPa}\sqrt{\text{m}}$ [16]; consequently, this means that the fracture strength of the dentin will be reduced by as much as a factor of two in the presence of a 300-μm sized crack, as compared to one ~100 μm in size. Clearly, laser treatments which can potentially generate such sized cracks should be avoided.

Cracking of teeth can therefore have serious clinical consequences which usually include the need for extensive

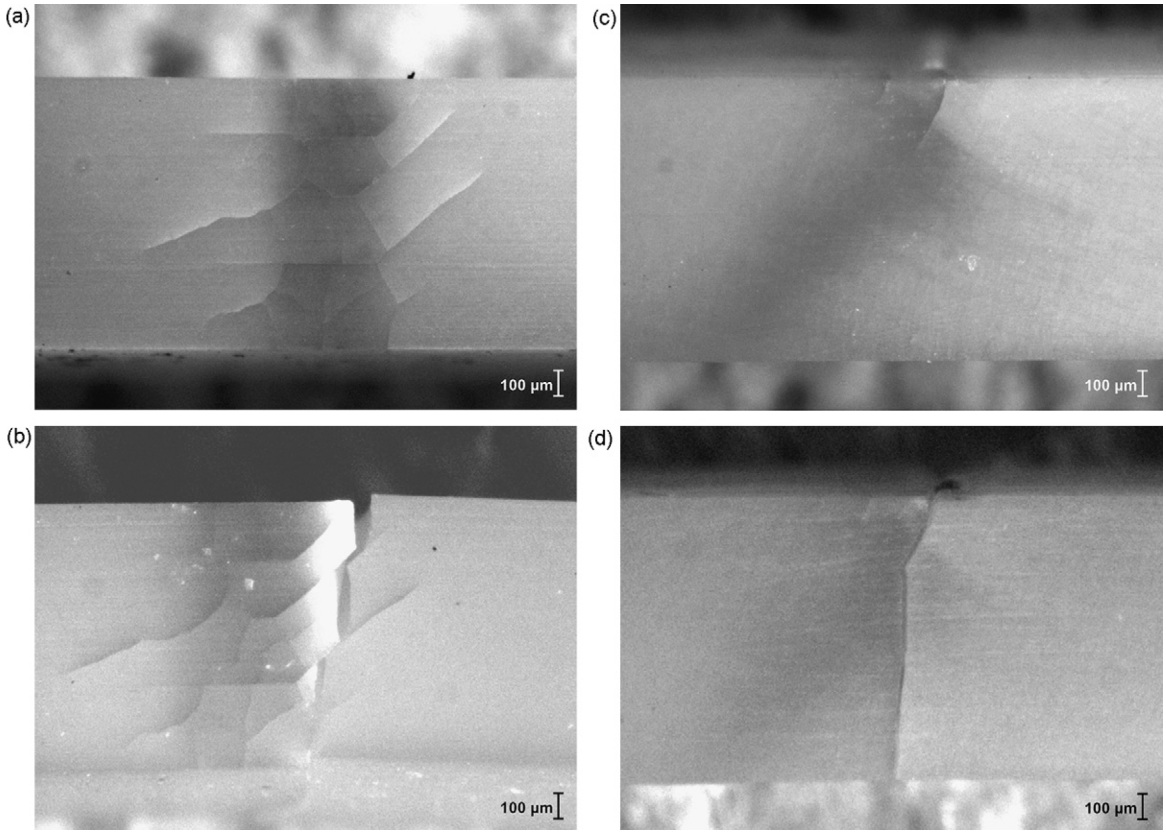


**Fig. 3 – (a) Irradiated surface, group D and (b) same sample, group D, after breaking. The failure is at the site of the initial cracks produced by irradiation. (c) Side view of a sample in group D, showing cracks on top surface and (d) side view of same sample after breaking, showing that the location of failure is a continuation of the initial crack produced by irradiation.**

restorative treatment [1,17,18]. Existing intra-coronal restorations predispose teeth to fracture, most often involving the loss of a cusp in a posterior tooth [19]. The internal form of an intra-coronal preparation may affect the susceptibility of the tooth to fracture at least in theory, because sharp angles tend to concentrate stresses [20,21]. If laser ablation is to be used for preparing intra-coronal cavity preparations, the quality of the surface could affect the subsequent susceptibility of the restored tooth to fracture. Even if the internal form of the resulting preparation is rounded, small surface cracks can still serve as initiation sites for eventual catastrophic failure. The vast majority of cavity preparations in clinical practice is done with handpieces. One previous study compared the generation of cracks by handpiece preparation to Er-YAG laser preparation; on SEM examination neither technique was found to generate cracks [22].

There are other considerations as well, such as the suitability of the surface for bonding procedures which are now commonly used for restoring teeth. Our previous study showed that different laser conditions result in surfaces with different adhesive properties for bonding composite resins. Thermal damage was particularly related to lower bond strength of the joints [23]. The laser conditions in group W in the present study were previously demonstrated to result in efficient ablation as well as good bonding characteristics without the need for etching.

In summary, Er-YAG laser irradiation with water and 35 μs pulse duration resulted in a surface without visible cracks but with a ~20% reduction in the bending strength of the dentin. Irradiation with a Er-YSGG laser in the absence of water, with 0.5 μs pulse durations, however, resulted in significant surface cracks which served as sites of initiation of catastrophic fractures, resulting in a 35% weakening of dentin under bending forces.

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## REFERENCES

[1] Bader JD, Martin JA, Shugars DA. Preliminary estimates of the incidence and consequences of tooth fracture. J Am Dent Assoc 1995;126:1650–4.

[2] Caron GA, Murchison DF, Cohen RB, Broome JC. Resistance to fracture of teeth with various preparations for amalgam. J Dent 1996;24:407–10.

[3] Kuroe T, Caputo AA, Ohata N, Itoh H. Biomechanical effects of cervical lesions and restoration on periodontally compromised teeth. Quintessence Int 2001;32:111–8.
[4] Kuroe T, Itoh H, Caputo AA, Konuma M. Biomechanics of cervical tooth structure lesions and their restoration. Quintessence Int 2000;31:267–74.
[5] Couegnat G, Fok SL, Cooper JE, Qualtrough AJ. Structural optimization of dental restorations using the principle of adaptive growth. Dent Mater 2006;22:3–12.
[6] 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;35:214–28.
[7] White JM, Goodis HE, Marshall SJ, Marshall GW. Sterilization of teeth by gamma radiation. J Dent Res 1994;73:1560–7.
[8] Staninec M, Gardner AK, Le CQ, Sarma AV, Fried D. Adhesion of composite to enamel and dentin surfaces irradiated by ir laser pulses of 0.5–35 micros duration. J Biomed Mater Res B: Appl Biomater 2006;79:193–201.
[9] Motamedi M, Rastegar S. Thermal stress distribution in laser irradiated hard dental tissue: implications for dental applications. Laser–tissue interaction III. SPIE 1992;1646:316–21.
[10] Vogel A, Venugopalan V. Mechanisms of pulsed laser ablation of biological tissues. Chem Rev 2003;103:577–644.
[11] Jacques SL. Role of tissue optics and pulse duration on tissue effects during high-power laser irradiation. Appl Opt 1993;32:2447–53.
[12] Lost C, Irion KM, John C, Nussle W. Two-dimensional distribution of sound velocity in ground sections of dentin. Endod Dent Traumatol 1992;8:215–8.
[13] Zuerlein MJ. Interactions of IR lasers with dental hard tissues at clinically relevant wavelengths. PhD Thesis. Graduate Group Bioengineering. University of California, San Francisco; 1999.
[14] Darling CL, Fried D. Real-time near IR (1310 nm) imaging of $CO_2$ laser ablation of enamel. Opt Express 2008;16:2685–93.
[15] Le CQ, Staninec M, Fried D. The influence of pulse duration on the bond strength of dentin to composite after Er:Yag laser irradiation. In: Proceedings of the SPIE, Lasers in Dentistry XI, vol. 5687. 2005. p. 151–6.
[16] Imbeni V, Nalla RK, Bosi C, Kinney JH, Ritchie RO. In vitro fracture toughness of human dentin. J Biomed Mater Res A 2003;66:1–9.
[17] Burke FJ. Tooth fracture in vivo and in vitro. J Dent 1992;20:131–9.
[18] Heft MW, Gilbert GH, Dolan TA, Foerster U. Restoration fractures, cusp fractures and root fragments in a diverse sample of adults: 24-month incidence. J Am Dent Assoc 2000;131:1459–64.
[19] Jagadish S, Yogesh BG. Fracture resistance of teeth with class 2 silver amalgam, posterior composite, and glass cermet restorations [published erratum appears in oper dent 1990 November–December, 15(6), 234]. Oper Dent; 1990, 15: 42–7.
[20] Johnson EW, Castaldi CR, Gau DJ, Wysocki GP. Stress pattern variations in operatively prepared human teeth, studied by three-dimensional photoelasticity. J Dent Res 1968;47:548–58.
[21] Kahler B, Kotousov A, Melkoumian N. On material choice and fracture susceptibility of restored teeth: an asymptotic stress analysis approach. Dent Mater 2006;22:1109–14.
[22] Sehy C, Drummond JL. Micro-cracking of tooth structure. Am J Dent 2004;17:378–80.
[23] Sheth KK, Staninec M, Sarma AV, Fried D. Selective targeting of protein, water, and mineral in dentin using UV and IR pulse lasers: the effect on the bond strength to composite restorative materials. Lasers Surg Med 2004;35:245–53.