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WATER INDUCED MECHANICAL EFFECT ON THE DENTAL HARD TISSUE BY THE SHORT PULSED LASER

YONG-HOON KWON[1*] and YOU-YOUNG KIM[2]
[1]Department of Physics, Kyungpook National University, Taegu 702-701, Korea
[2]Department of Biochemistry, Kyungpook National University, Taegu 702-701, Korea

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Abstract – One macroscopic effect in the free-running Er:YAG laser is an accumulation of microscopic effects. Understanding of the exogenous water induced mechanical effect on the dental hard tissue by the Q-switched Er:YAG laser has an important impact on the further understanding of the free-running Er:YAG laser ablation on the dental hard tissue. The Q-switched Er:YAG laser (1-μs-long pulse width) was used in the recoil pressure measurement with an aid of water-jet system and a pressure transducer. The amplitude of the recoil pressure depends on the tooth surface conditions (dry and wet) and the volume of the water upon it. Wet surfaces yielded higher recoil pressure than that of dry surface, and as the volume of the exogenous water drop increased, the amplitude of the recoil pressure increased also.

INTRODUCTION

Laser ablation of dental hard tissue has been the subject of researches ever since the ruby laser was studied for its possible dental applications. Stern and Sognnaes[1] showed that the ruby laser has the ability to ablate hard tissue. However, the laser caused severe, irreversible damage to the tooth. Since then, various mode of lasers including CW lasers and free-running and Q-switched pulsed lasers[2-10] have been tested on dental hard tissue to determine the plausibility of the laser as a controllable dental handpiece. For the laser to be useful for dental treatment, it should satisfy two criteria: (1) it should have a better drilling depth and (2) it should cause less mechanical and thermal damage to surrounding tissue of the crater than a mechanical dental handpiece. For these criteria, most researchers have been focused on exploring the ability of pulsed lasers to remove dental hard tissue in a safe and efficient manner. Currently available medical lasers usually generate optical radiation in the visible and near-IR ranges of the spectrum where hard tissues have very weak light absorption.[11,12] However recent research on the application of dyes in combination with pulsed lasers could solve the problem of poor laser absorption in the hard tissue, and significantly improves ablation in the near-IR region[13,14] becuase of their similar absorption band peak.

According to the research to date, the Er:YAG laser is the most promising for dental hard tissue treatment.[15] This is becuase water, which is a minor constituent of hard tissue, shows the highest absorption at the Er:YAG laser wavelength.[16] In the study to circumvent the scarcity of water content in the tooth, an exogenous water drop was put on the surface of the tooth. Actually water is already used in clinical dentistry to cool a tooth during laser treatment.[17,18] Adding an exogenous water drop on the tooth is similar to applying dyes on the tooth in order to promote the laser light absorption.[19] The advantagse of the water drop method are that (1) water is harmless, (2) a repeatedly dispensed exogenous-water-drop on the tooth lowers the temperature rise in the tooth, and (3) a Microjet system can precisely control the volume and frequency of a water drop jet. All these are important parameters for the optimal dental hard tissue ablation.

Understanding the Q-switched Er:YAG laser (1-μs-long) induced dental hard tissue ablation is important for understanding the mechanical effect which occurs in the free-running Er:YAG laser mode because one macro pulse of the Er:YAG laser is composed of about 25 1-μs-long micro pulses.

The objective of this work was to investigate the effects of exogenous water on dental hard tissue ablation in the Q-switched laser. Studies focused on measuring, using a calibrated transducer, the absolute amplitude of recoil pressure waves induced during the ablation of enamel and dentin.

MATERIALS AND METHODS

A Q-switched Er:YAG laser (Schwartz Electro-Optics, wavelength = 2.94 μm) was used in the experiments. The laser had a 2 Hz repetition rate and a 1 μs pulse duration. Each laser pulse profile was detected and recorded by an indium arsenide photodetector (J12-LD2-R250U) and a digital scope (Tektronics, TDS320), respectively. Incident laser pulse energies were measured with a joulemeter (Molectron,

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* To whom correspondence should be addressed.

JD2000). Measured energies ranged from 20 mJ to 55 mJ. Focused laser spot diameters was 500 μm. Extracted healthy human molars were sectioned along the longitudinal direction approximately 1 mm thick using a slow-speed diamond saw. Small droplets of distilled water were dispensed using a Microjet system on the surface of the tooth to enhance laser light absorption. A Microjet system is a piezoelectric ink-jet device which can precisely dispense pre-set amount of water from 400 pL to 400 nL.

A calibrated broad-band (0.5-40 MHz, 25 ns temporal resolution) piezoelectric transducer was employed to measure the transient temporal profile and the absolute values of laser-induced recoil pressure on the irradiated zone. Simultaneously detected transient temporal profile of the pressure waves and Q-switched Er:YAG laser pulse were displayed on the digital scope. Both transducer and photodetector signals were displayed and stored on the connected computer.

In the experiments, three different enamel and dentin surface conditions were employed: dry, water drop, and water layer. Volumes of the water drop were 20 nL and 400 nL. The thicknesses of the water layers were 300-400 μm. Generated pressure waves were propagated through the layer in two ways:

1. water drop or water layer →enamel(1 mm)→thin water layer between the sample and acoustic transducer (several micron)→ aluminum conductor(3 mm thickness) → piezoelectric detector.
2. enamel →thin water layer between the sample and acoustic transducer (several micron)→aluminum conductor (3 mm thickness)→piezoelectric detector.

Once pressure waves generate in the irradiated zone, they propagate along the laser beam axis in the sample in two direction (some of them into the sample, and the rest toward the sample surface). Transmittance of pressure waves through a boundary of different acoustic impedances with a condition of perpendicular propagation of waves to the surface of an interface can be expressed as

$$T = \frac{Pt}{Pi} = \frac{2Z_A}{Z_A + Z_B} \qquad (1)$$

where impedance ($Z = \rho \cdot c$ = medium density speed of sound in the medium), and $Z_A$and Z_B are associated with Ahead of and Behind of propagating waves respectively. From the formula (1), we can see the increase of transmittance in accordance with the impedance mismatch at the boundary. If the impedance of Ahead is larger than that of Behind, transmittance of pressure waves will be enhanced; otherwise it will be minimized.

Absorption and scattering of the acoustic wave energy cause the generated pressure waves to attenuate as they propagate deeper into the medium. In the frequency range of interest (f ~ 1 MHz) the acoustic attenuation in water and aluminum is insignificant because of the very low absorption coefficient of water and aluminum.[20] The acoustic attenuation coefficients measured in enamel and dentin are very limited in number and are not consistent with each other.[21,22] Within the limit of published data,[23] ultrasonic attenuation in dentin at 1 MHz can be calculated by extrapolating given data and then by assuming data's linear dependance on the acoustic frequency. By doing that we could get the attenuation coefficient α_{dentin} = 7 NP/cm for dentin. Based on this coefficient and by assuming exponential decrease of recoil pressure amplitude in the medium, the amplitude of the attenuating recoil pressure at the bottom of 1 mm thick dentin becomes one half of the initial recoil pressure.

The finite size of an area of stress generation causes the diffraction of pressure waves such that a plane wave gradually transforms into a spherical wave as it propagates. Spherical waves have a much wider effective area of acoustic wave than plane waves do, and as a result pressure amplitude decreases. The diffraction process is important for low acoustic frequencies and small beam diameters. To ensure that the stress waves remain plana during their transit across the sample, larger laser spots are required. The large spot size also ensures that the expansion of ablated material away from the target surface will be of a one-dimensional geometry. The ratio of acoustic area at the sample to the transducer surface is expressed by $A_{\text{Laser}}/A_{\text{Transducer}}$. By taking into account attenuation, transmittance, and diffraction through boundaries and medium we can calculate the pressure amplitude at the surface of dentin(enamel) from the measured pressure signal at the bottom of dentin(enamel) and the transducer sensitivity.

RESULTS AND DISCUSSION

Studies of the ablation values of recoil pressure waves and the transient history of the pressure profile upon the ablation of enamel and dentin with Q-switched laser pulses are essential for understanding laser-induced mechanical effects on hard tissue ablation. Transient pressure wave profiles with different surface conditions of dry, water drop, and water layer for enamel are presented in Figs. 1 and 2.

A monopolar nature of transient stress waves proves that the pressure profile we measured was produced by the ablation process. The first peak pressure waves on enamel with water drop and water layer were due to the ablation of exogenous water. The dip in the middle of the stress profile is due to the reflection of the initial compression wave from the air/water interface for the water-wet surface, or from the air/enamel interface for the dry surface. The time delay between the moment of laser irradiation and the propagated acoustic pulse detection was measured and eliminated in the course of data processing in order to compare the amplitude of recoil presssures. Exogenous water, which initially was dispensed to enhance the absorption of laser energy, has enhanced the amplitude of recoil pressue also. In Figs. 3 and 4, we could observe the same enhancement of the pressure though the sample has changed from enamel to dentin.

Differences in the physical and mechanical properties

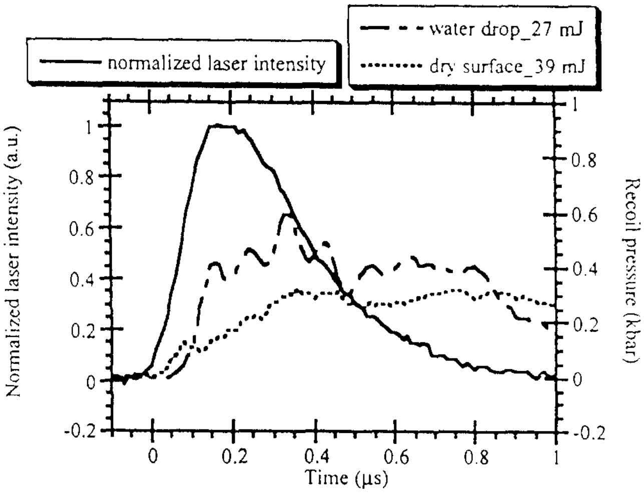

Figure 1. The profiles of transient pressure generated upon enamel ablation with dry and exogenous water drop on the surface.

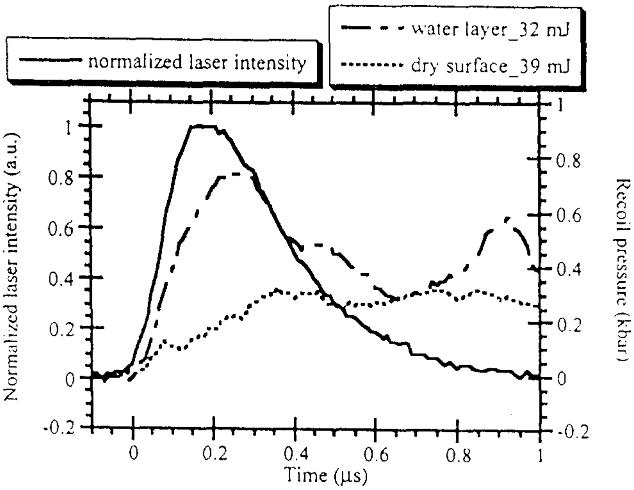

Figure 2. The profiles of transient pressure generated upon enamel ablation with dry and water layer on the surface. The thicknesses of water layer were about 300 - 400 μm.

of enamel and dentin influence the acoustic wave velocity and acoustic wave attenuation in the medium. Note in Fig. 5 that the differences in enamel and dentin influence the profile of stress waves. The acoustic wave velocity in enamel (6000 m/s) is much faster than it is in dentin (4000 m/s). This means that, given the same surface condition and laser energy, the recoil pressure waves in enamel will arrive earlier at the transducer such that the rising slope in the pressure profile appears first in enamel and then in dentin. The amount of acoustic wave attenuation data is very limited, and the reported data shows large variance. However, the difference of attenuation coefficients between enamel and dentin is apparent because the measured amplitudes of recoil pressure waves in enamel and dentin with the same surface conditions and laser energy showed significant differences in their magnitude. Since the attenuation of acoustic waves in enamel is much higher than it is in dentin, the amplitude of recoil pressure waves in enamel was lower than it was in dentin in all surface conditions with the same laser energy.

The temporal profiles of transient recoil pressure waves induced by the Q-switched laser ablation of enamel and dentin with and without exogenous water on the surface disclose important features of the effect of water on the ablation process. Monopolar nature in figures tells that the recoil pressure is induced by a strong ablation. Once stress waves generate from the laser irradiated zone, some of them transmit into the medium and the rest reflect from the interface. Similar

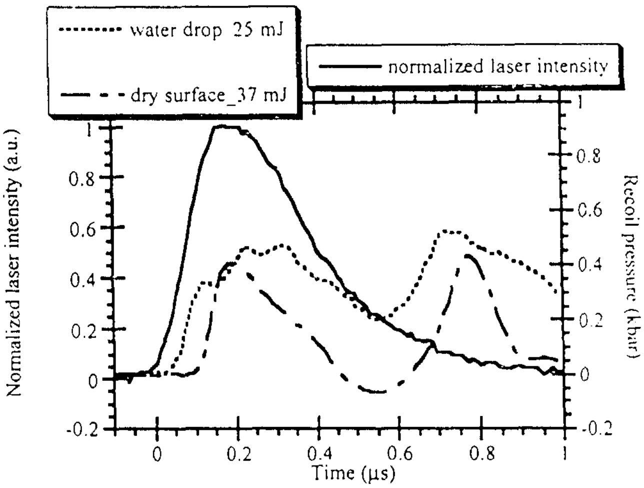

Figure 3. The profiles of transient pressure generated upon dentin ablation with dry and exogenous water drop on the surface.

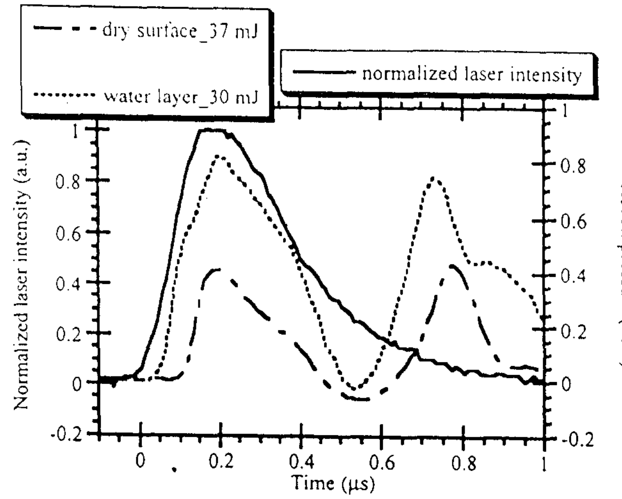

Figure 4. The profiles of transient pressure generated upon dentin ablation with dry and water layer on the surface. The thicknesses of water layer were about 300 - 400 μm.

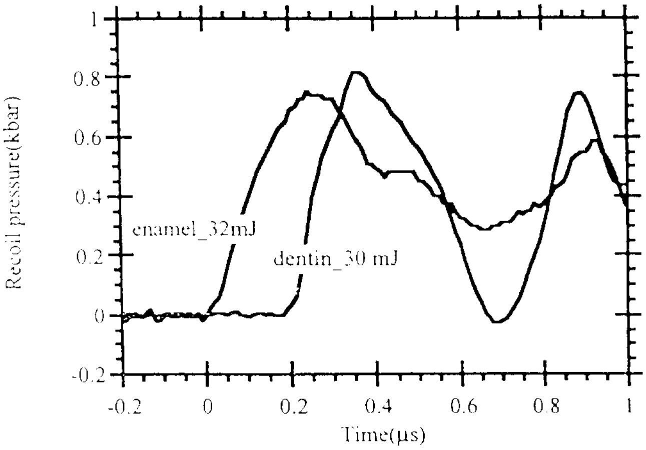

Figure 5. The Q-switched laser-induced profile of transient pressure waves generated upon dentin and enamel. The differences in the acoustic wave velocity and attenuation coefficient in the tooth affect the initiation of the rising slope and the peak amplitude of recoil pressure in the tooth.

to formular (1), we can define reflectance R as follows:

$$R = \frac{Pr}{Pi} = \frac{Z_A - Z_B}{Z_A + Z_B} \qquad (2)$$

In a rigid boundary, Z_A is the impedance of water (1.5 × 105 g/cm^2 s), and in a free boundary, Z_A is the impedance of air (40 g/cm^2 s). The estimated compressive pressure in dry surface enamel is in the lower limit of ultimate compressive strength of enamel which ranged between 950 bar to 4 Kbar.[23] The estimated tensile strength in dry enamel is, given the same order of initial pressure, higher than the ultimate tensile strength which ranged 300-350 bar.[23] Since the amplitude of stress waves in our experiment is always higher than the threshold for mechanical failure of enamel, ablation occurs every time.

The magnitude of the recoil pressure waves generated by the ablation depends on water for two reasons. The first reason is the form of water itself. Water drop or water layer on the tooth changes the surface condition from a free boundary to a rigid boundary. Transmittance of the generated initial pressure in a rigid boundary (water-enamel interface) is 1.85 times greater than that of a free boundary (air-enamel interface). Enhancement of the generated recoil pressure was accomplished just by changing the surface condition. The second reason involves the ablation of water. Recoil force always accompanies the ablation of material because of the conservation of momentum. The volume of explosively evaporating water is dependent on the volume of exogenous water on the enamel surface. In basic physics, the recoil pressure is defined as (recoil force / enforced area). Recoil force is the temporal integration of recoil momentum. Recoil momentum is the product of total mass lost in the medium and the average velocity of ejecting materials. From these physical relationships, it is clear that the total mass loss of water depends on the total volume of exogenous water such that the greater the volume of ablated water, the stronger the recoil pressure waves on enamel will be. However, increasing the volume of water on the enamel surface requires an increase of laser energy for water ablation, and will result in decreased remaining laser energy for enamel ablation.

CONCLUSION

The study of the exogenous water induced dental hard tissue ablation with the Q-switched Er:YAG laser is important for the further understanding of the one macroscopic effect which occurs in the hard tissue ablation by the free running Er:YAG laser. The effect of water in the hard tissue ablation was studied via the recoil pressure analysis for different tooth surface conditions: dry surface and surface with exogenous water. The amplitude of the recoil pressure on the tooth was increased as the volume of exogenous water increased. In every case, the amplitude of recoil pressures with exogenous water was higher than the ultimate tensile strength of the enamel (300-350 bar). The mechnical effect on the free-running Er:YAG laser is an accumulation of the effects induced by many micropulses which have the same pulse width we have employed in this Q-switched Er:YAG laser study. For this reason though the Q-switched Er:YAG laser itself is not an efficient hard tissue drill, it is useful for the analysis of the free-running Er:YAG laser ablation mechanism. The results we found will be useful for the determination of the optimal and safe ablation conditions with exogenous water in the free-running Er:YAG laser.

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