Pulpal Temperature Changes during Low-power Hard-tissue CO_2 Laser Procedures

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Thermal insult to pulpal tissue is recognized as a major limitation to the use of lasers for dental hard-tissue procedures. This study examined thermal changes at the level of the dental pulp in human molar teeth irradiated with a CO_2 dental laser using a pulsed mode of operation. Sectioned molar teeth were exposed, in vitro, to CO_2 laser radiation. The laser parameters were those used clinically for laser desensitization and laser-enhanced fluoride treatment. Fissure regions and root surfaces were irradiated. For settings which might reasonably be used clinically, the temperature rise was not of a magnitude which would be expected to cause pulpal inflammation or necrosis. With regard to thermal properties of tooth structure, times taken to reach the maximum temperature reduced, and times taken to cool to baseline increased with increasing laser exposures.

Key Words: lasers, dental pulp, thermal stress.

Introduction

Carbon dioxide (CO_2) lasers have attracted interest in recent years for use in various dental hard-tissue procedures. CO_2 laser radiation is within the infra-red region of the electromagnetic spectrum, with a wavelength of 10.6 mm. This laser wavelength is absorbed strongly in dental enamel, dentin and cementum (Featherstone and Nelson, 1987). Consequently, useful working temperatures of hundreds to thousands of degrees Celsius may be generated at the laser impact site, with, theoretically, minimal transmission of thermal energy into the pulp (Fowler and Kuroda, 1986). Indeed, the temperature rise in the pulp chamber/root canal has been shown, for the continuous wave mode of emission, to be minimal (Powell et al., 1989). This likely reflects the inherent thermal conductivity of dentin. Parameters for thermal injury to dental pulp have been defined by Zach and Cohen (1965), who examined histological changes in pulps of monkeys exposed to a non-laser heat source (soldering iron). Their work defined a critical threshold for temperature rise of 10oF (5.5oC), above which an unacceptably high incidence of pulpal necrosis occurred. Below 5.5oC, reversible and mild pulpitis occurred. Below 4oF (2.2°C), no histological changes were discernible. Based on these threshold values, subsequent studies using continuous wave CO_2 lasers as a heat source have concluded that exposures should not exceed 10 Joules (J), to ensure that the critical temperature increase of 5.5°C does not occur (Miserendino et al., 1989; Powell et al., 1989). These same studies determined that, to avoid pulpal necrosis, continuous wave exposures should not exceed 30 J. However, there are no data available on the thermal effects of CO_2 lasers operated in pulsed (chopped) modes. From a purely theoretical standpoint, pulsed modes of operation are safer, in that cooling can occur between laser pulses (Walsh, 1993). The present study was undertaken to quantify temperature changes during lasing of crown and root regions using pulsed lasing modes identical to those employed for laser desensitization and enhanced fluoride uptake. The study had two aims. Firstly, to determine which settings fell within the pulpal safety limits defined by Zach and Cohen (1965), and secondly, to examine what differences existed between temperature rises with different pulsed modes. The times taken for the maximum temperature to be reached, as well as the times to return to baseline temperature (thermal relaxation) were also examined.

Material and Methods

A Luxar LX-20 dental laser was used (Luxar Corporation, Bothell, Washington, USA). The beam was delivered via a flexible hollow waveguide to which a straight handpiece with a ceramic tip was attached. The laser spot size was 0.8 mm. The laser power output was monitored, and the delivery system was calibrated twice daily. The laser power was maintained at 2 Watts. The energy dose (E) was calculated by the following formula: E = Power (W) x duty cycle x time (s), where duty cycle (DC) = pulses per second (Hz) x length of pulse (s) x 100%. Continuous wave mode corresponds to a duty cycle of 100%. Surgically removed unerupted human third molar teeth were sectioned using an Isomet diamond saw. Sections were cut either horizontally (crown segments) or vertically (root segments) to achieve a consistent thickness (crown segments 2.0 mm, root segments 2.5 mm), measured from the tooth surface to the pulp chamber. Sectioning was performed so that the thermocouple could be placed in the pulpal site at greatest risk of thermal insult. The pulp contents were removed, and the tooth sections mounted in plaster discs. Specimens were kept moist at all times in order to simulate intra-oral conditions of humidity. The method used to assess temperature change at the dental pulp has been described in detail previously (Sandford and Walsh, 1994). In brief, a K-type bead thermocouple was located securely against the pulpal tooth surface by means of non-heat-conductive plaster weights. The thermocouple was connected in turn to a temperature-to-voltage converter, which provided an output of 1 mV per degree Celsius. The voltage output was measured with a digital multimeter which provided a digitally encoded output to the serial interface of an IBM-compatible computer. Data for temperature were recorded at intervals of five seconds. After establishing the baseline temperature, specimens were lased within the fissure region or coronal root surface, directly opposite the thermocouple bead. The laser was held at a constant distance of 1.5 mm from the target site. Recordings were made continuously during and after lasing, until the samples had cooled to baseline temperature. Each trial was replicated a minimum of five times. The following parameters were evaluated: (i) temperature rise (difference between baseline temperature, and the maximum temperature recorded during the trial); (ii) time to maximum temperature (calculated from the start of lasing); and (iii) time to cool to baseline (calculated from the timepoint at which the maximum temperature occurred). Differences were evaluated using two-way paired analysis of variance.

Results

Temperature rise - Crown

As shown in Figure 1, the temperature following lasing of the crown increased with increasing laser exposure. However, temperature rises for all but one of the settings (60 seconds/5% DC/6.0 J) were below the pulpal injury threshold of 2.2°C. Differences between the 1% and 5% duty cycles were significant for irradiation times of 5, 10, and 30 seconds ($P<0.05$), but not for 60 seconds.

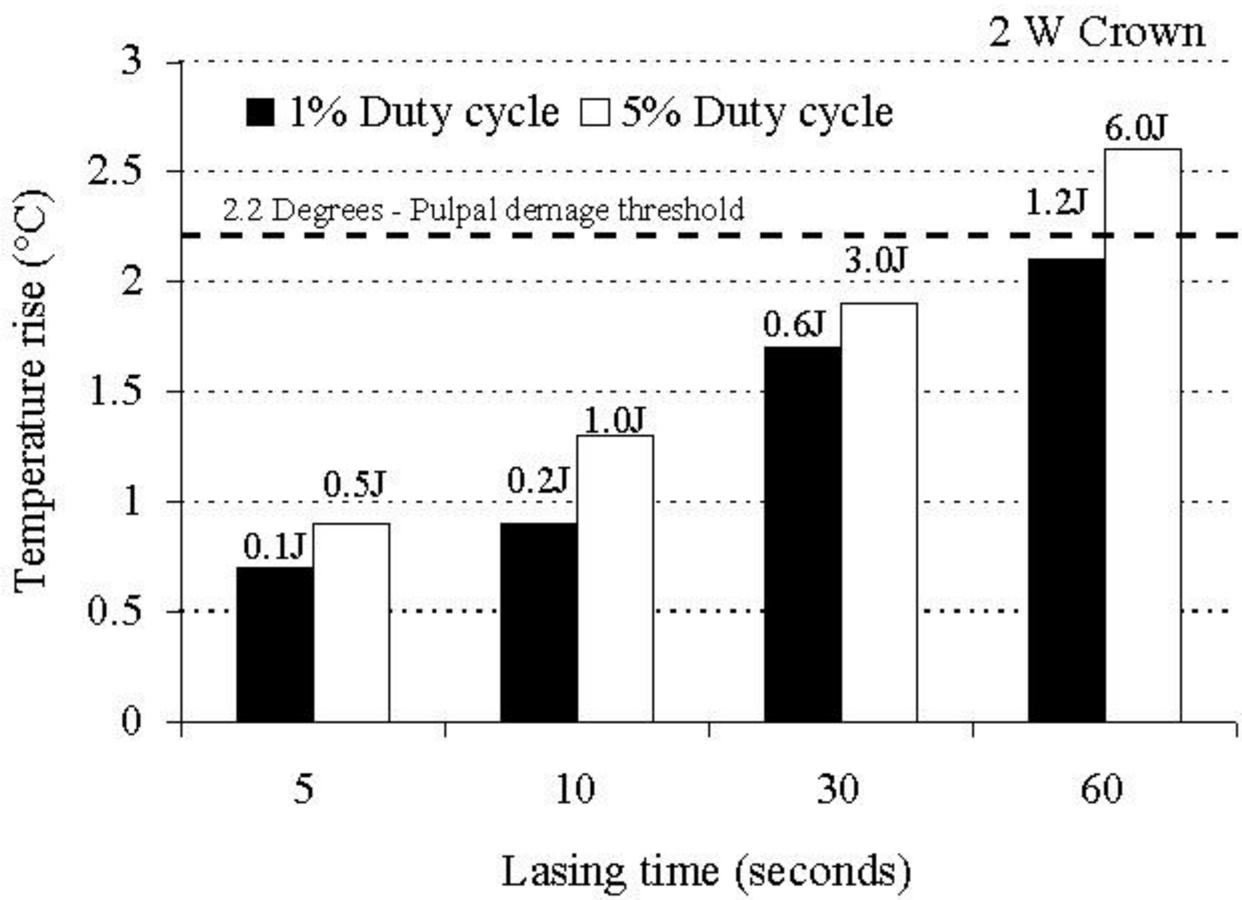
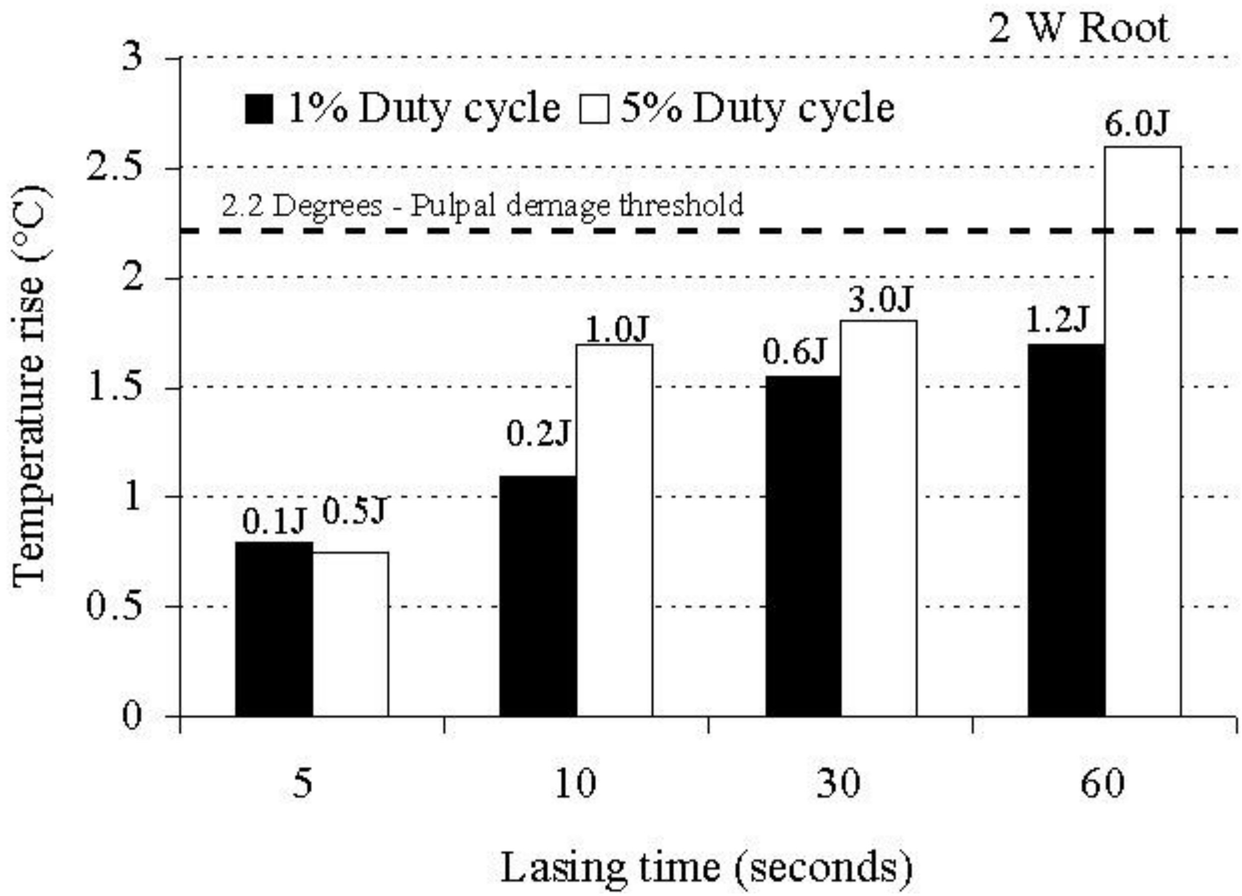


Figure 1 - Temperature rise following irradiation of the crown region. Bars show mean values (N = 5 replicates).

Temperature rise - Root

As shown in Figure 2, temperature rises with the dental pulp for laser irradiation of root surfaces increased in proportion to the laser exposure. Again, as with the crown series, temperature rises for all but one of the settings (60 seconds/5% DC/6.0 J) were below the threshold value of 2.2°C. Differences between the 1% and 5% duty cycles were significant for lasing times of 10, 30, and 60 seconds ($P<0.01$, $P<0.01$, and $P<0.05$ respectively). Values for trials of 5 seconds laser exposure were not significantly different. There was a linear relationship between laser exposure and pulp temperature rise for each of the laser parameters investigated in the study. Figure 3 shows linear regression analysis, with values for gradients (m) and correlation coefficients (r).



[Figure 2](#) - Temperature rise following irradiation of the root surface. Bars show mean values (N = 5 replicates).

Time to reach maximum temperature

The times taken to reach maximum temperature were related to the duration of laser exposure, with the maximum temperature occurring within 10 seconds of completion of lasing in all instances. Data for these times are presented in [Table 1](#). Each value is the mean of five readings, rounded to the nearest second.

Table 1 - Time to reach maximum temperature (in seconds).

	Root				Crown			
	5 s	10 s	30 s	60 s	5 s	10 s	30 s	60 s
1% Duty cycle	12	15	34	66	6.8	10	30	55
5% Duty cycle	10	11	30	56	15	17	35	64

Time (in seconds) from the commencement of lasing to the temperature peak (values are the mean of 5 experiments).

Table 1.

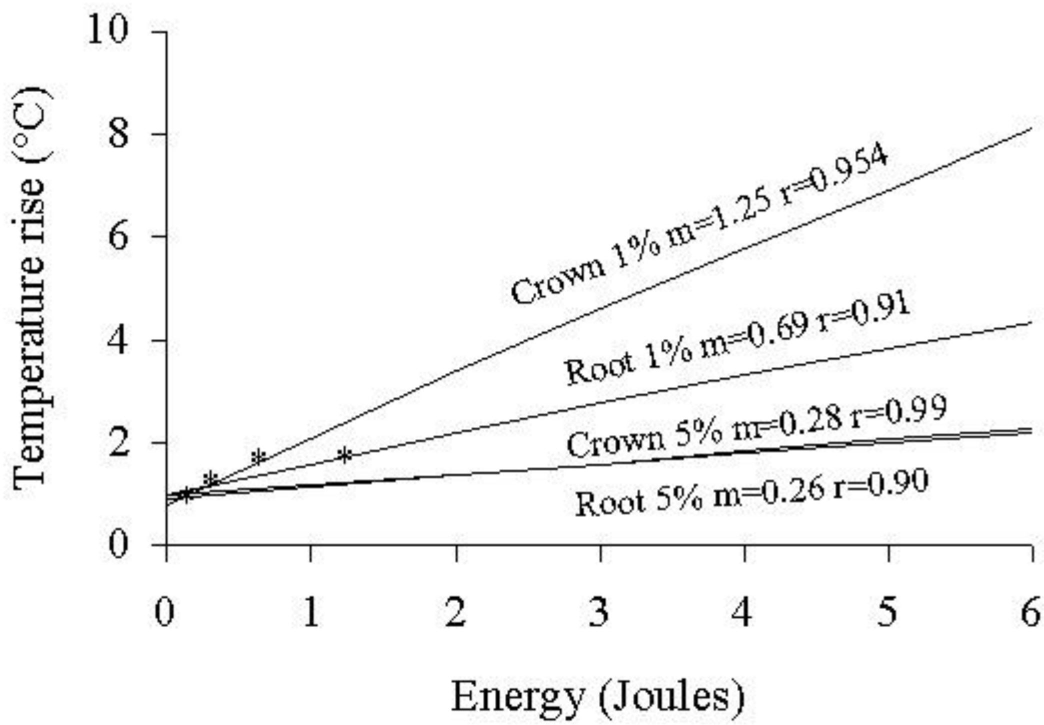


Figure 3 - Linear regression analysis of the relationship between laser exposure (Joules) and temperature rise.

Time to cool to baseline
Cooling times were related to the duration of laser exposure, with longer lasing periods associated with longer cooling times. Data for times are presented in **Table 2**.

Table 2 - Time to cool to baseline temperature (in seconds).

	Root				Crown			
	5 s	10 s	30 s	60 s	5 s	10 s	30 s	60 s
1% Duty cycle	45	62	77	74	32	40	51	67
5% Duty cycle	48	60	76	100	37	44	65	75

Time (in seconds) from the start of lasing until the sample had cooled to the baseline temperature (values are the mean of 5 experiments).

Table 2.

Discussion
This study examined thermal changes in teeth following exposure to CO_2 lasers operated in pulsed mode. The results indicate that the combination of low power and low duty cycle induces minimal thermal changes at the level of the dental pulp. Temperature rises for the 1% duty cycle were less than those for the 5% duty cycle, and this reflects the increased opportunity for cooling to occur between laser pulses. The 5% duty cycle comprises ten 5 millisecond exposures per second, with 950 milliseconds per second for cooling (95 milliseconds after each laser pulse). In contrast, the 1% duty cycle setting comprises two 5 millisecond exposures per second, allowing 990 milliseconds per second for cooling (495 seconds after each laser pulse). It should be noted that the two instances in which differences between these two modes were not significant reflect proportionally high standard deviations in the sample groups. Significant differences may be obtained with larger sample sizes. With pulsed modes of lasing, the maximum temperature is always reached at or immediately after the end of lasing. This may be attributed to the thermal conductivity of the dentin and enamel/cementum, in that the thermal pulse dissipated within the tooth takes several seconds to reach the thermocouple bead. Times to maximum temperature may be even longer in the mouth than those recorded in this study, due to cooling from pulpal blood flow and saliva. These same two factors would, similarly, be expected to reduce cooling times. This study demonstrates that CO_2 lasers, even at low duty cycles, invariably induce a temperature rise within dental pulp during lasing in vitro of the crown or root surface at settings used for desensitization (Forrest-Winchester and Walsh, 1992) or enhanced fluoride treatment (Walsh, 1993). Only the highest of the energy settings

used in the study (6.0 J) produced temperature changes (both in crown and root) which were greater than the 2.2°C pulpal damage threshold defined by Zach and Cohen (1965). An irradiance of 6.0 J is clinically unrealistic for these hard-tissue procedures. Thermal changes at low energy settings are unlikely to be significant at the clinical level, as pulp symptomatology and other adverse changes have not been observed (Walsh, 1994). Thus, pulsed lasing modes are recommended strongly for dental hard-tissue procedures because they provide the opportunity for cooling between exposure pulses.

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References

Featherstone JDB, Nelson DGA: Laser effects on dental hard tissues. Adv Dent Res 1: 21-26, 1987

Fowler BO, Kuroda S: Changes in heated and in laser-irradiated human tooth enamel and their probable effects on solubility. Calcif Tiss Int 38: 197-208, 1986

Forrest-Winchester K, Walsh LJ: The effect of infrared laser radiation on dentinal permeability in vitro. Periodontol 3: 37-43, 1992

Miserendino LJ, Neiburger EJ, Walia H, Luebke N: Thermal effects of continuous wave CO2 laser exposure on human teeth: an in vitro study. J Endodont 15: 302-305, 1989

Powell GL, Morton TH, Larsen AE: Pulpal response to irradiation of enamel with continuous wave CO2 laser. J Endodont 15: 581-583, 1989

Sandford MA, Walsh LJ: Differential effects of pulsed vs. continuous CO2 laser radiation on human molar teeth. J Clin Laser Surg Med 12: 139-142, 1994

Walsh LJ: Pulpal safety parameters for irradiation of dental hard tissues with carbon dioxide lasers. Aust Endodont Newsl 19: 21-25, 1993

Walsh LJ: Clinical evaluation of dental hard tissue applications of carbon dioxide lasers. J Clin Laser Med Surg 12: 11-15, 1994

Zach L, Cohen G: Pulp responses to externally applied heat. Oral Surg Oral Med Oral Pathol 19: 515-530, 1965

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