Thermal Changes in the Dental Pulp During Er,Cr:YSGG Laser Removal of IPS e.max Press Lithium Disilicate Veneers

Thesis

Presented in Partial Fulfillment of the Requirements for the Degree of Masters of Science in The Graduate School of The Ohio State University

By

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2012

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Abstract

**Background:** Lasers have become an increasingly popular tool in the field of dentistry due their ability to perform a wide variety of both hard and soft tissue procedures. Recently it has been proposed that lasers may prove to be useful in the removal of some all-ceramic restorations, particularly veneers. The amount of temperature being generated intrapulpally during such a procedure has yet to be studied.

**Methods:** Thirty IPS e.max® Press lithium disilicate veneers were cemented to thirty extracted human premolars with Variolink® Veneer resin cement. Veneers were randomly divided into one of five groups, and scanned with an Er,Cr:YSGG laser at either 0W/0Hz, 25W/25Hz, 35W/25Hz, 25W/35Hz, or 35W/35Hz. During laser scanning, intrapulpal temperatures and debonding times were monitored.

**Results:** Increasing the laser wattage and/or pulse repetition rate resulted in an increase in the temperature generated intrapulpally. Increasing the wattage and/or
decreasing the pulse repetition rate resulted in a reduced debonding time. At both 0W/0Hz and 23W/35Hz veneers we unable to be debonded.

**Conclusions:** The laser group at 2.5W/25Hz was the overall safest laser group studied. Based upon the results of this study, laser removal of ceramic veneers should be a safe procedure for the dental pulp if the correct laser settings are used.
Dedication

Dedicated to my loving and patient wife, Claire
Acknowledgements

I would like to give special thanks to Dr. William Johnston for his valuable help with the development and interpretation of the statistics for the study. I would also like to thank all of my committee members for the generous sharing of their knowledge, experience and time, which contributed to making the study a success.
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Fields of Study

Major Field: Dentistry
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Chapter 1: Introduction

The Er,Cr:YSGG Laser

The erbium, chromium: yttrium-selenium-gallium-garnet (Er,Cr:YSGG) laser, is becoming an increasingly popular instrument in the field of dentistry since it can be used for both hard and soft tissue procedures including, but not limited to: caries removal, enamel etching, cavity preparation, root canal sterilization, crown lengthening, troughing, and blood coagulation. Advantages of the Er,Cr:YSGG laser include its ability to perform a variety of dental procedures without the aid of local anesthesia, the ability to selectively target specific tissues for ablation, minimal damage to adjacent hard and soft tissues during cutting, no vibration during cutting, and a minimal risk of thermal insult to the pulp (if used correctly). Disadvantages include the initial investment cost necessary to purchase the laser, as well as the potential to thermally damage the surface of the tooth or the dental pulp, if used incorrectly, which could lead to insufficient bonding of restorative materials or pulpal necrosis, respectively.

The Er,Cr:YSGG laser operates at a pulsed wavelength of 2.78µm and can therefore deliver its energy to the handpiece via a fiber optic cable. The optical power conduit between the handpiece and the target tissue is a sapphire or quartz
tip which should be immersed in an adjustable air-water spray during most procedures for both temperature and ablation control. The penetration depth in hard tissues is shallow because the Er,Cr:YSGG laser operates at a wavelength that is highly absorbed by water which is present on the surfaces of most tissues.

**Mechanism of Laser Ablation**

Ablation, as it is generally defined when referring to lasers in dentistry, describes the process by which laser energy of a given wavelength is used to target and surgically remove hard or soft dental tissue. The terms most commonly used in the literature to describe the process by which the laser accomplishes the ablation of oral structures include: *hydrokinetic system (HKS)*, photothermal effect, and *thermomechanical process*. There has been much speculation about the actual mechanism of tissue removal by lasers; however, these terms would lead one to assume that through the absorption of energy, and the generation of heat, tooth structure is mechanically removed.

As a review of the composition and structure of the hard tissues of the tooth; enamel, the outer protective layer of the tooth and hardest biological tissue in the body, is composed of approximately 96% inorganic matrix and 4% organic matrix/water by weight. The inorganic matrix is composed primarily of
hydroxyapatite ($\text{Ca}_{10}\text{[PO}_4\text{]}_6\text{[OH]}_2$) crystals arranged into a series of rods that extend from the surface of the tooth to the dentinoenamel junction. The organic matrix is made up of the protein enamelin, and water$^1$.

Dentin, the supporting hard tissue beneath enamel, is composed of 70% inorganic matrix, 20% organic matrix, and 10% water by weight.$^{1,55}$ Similar to enamel, the inorganic portion is formed primarily by hydroxyapatite, but unlike enamel, organic matrix is composed of collagen. Dentin components are arranged into tubules that extend from the dentinoenamel junction to the pulp. The tubules contain living odontoblasts that originally formed the tubules, and are thought to play a part in pain sensitivity$^1$.

The peak wavelength of absorption for water is $2.94\mu m$, and the peak absorption wavelength of hydroxyapatite is $9.6\mu m$. Due to the fact that both enamel and dentin are made up in large part by water and hydroxyapatite, the absorption spectrum of both tissue types exhibit strong absorption bands in the 2.7-3.0$\mu m$ and 9-11$\mu m$ range.$^{18,28,43}$ Many lasers available today for dental use (Er:YSGG, Er,Cr:YSGG, Er:YAG, and $\text{CO}_2$) operate at wavelengths within these spectrums for this reason.$^{36}$
The Er,Cr:YSGG laser used in this study emits a pulse beam of energy at a wavelength of 2.78µm (2,780nm).\textsuperscript{10,16,48} This wavelength, which falls into the mid-infrared (IR) range, is strongly absorbed by water, and the hydroxyl (-OH) groups of hydroxyapatite.\textsuperscript{9,23,25,28,31,36} It is believed that when laser energy comes into contact with enamel or dentin, it is absorbed by the water and hydroxyl groups, and the energy causes the water to be heated to extremely high temperatures, leading to its vaporization into steam, and the subsequent build-up of enormous pressure within the porous structure of the tooth. It is thought that as the pressure of the vaporized water molecules increase, it eventually surpasses the fracture strength of the remaining inorganic apatite material, and results in a series of microexplosions which mechanically separates and disperses the weakened hard tissue fragments.\textsuperscript{9,11,18,48}

**Ablation Rate**

The rate of ablation, which describes how effective the laser is during tissue removal, is highly influenced by water spray, pulse energy (W), pulse rate/frequency (Hz),\textsuperscript{31} tip distance from the target tissue (1-2mm),\textsuperscript{13,16,44} tip angulation, and tissue type, just to name a few. It is important to this study that basic understandings of how these settings influence the ablation rate of lasers be discussed.
The necessity of an exogenous water source to bathe the tip of the laser, and the target tissue, cannot be overemphasized. Ablative studies indicate that lasers used with water spray exhibit significantly increased cutting efficiency when compared to lasers use without water spray. The reasoning behind this observation, is that once the small amount of water present in the dental hard tissues is vaporized, the laser then requires an external source of moisture to continue to ablate effectively. The influence of the endogenous water present in the tissue is actually thought to have little impact on the ablative process, it is the exogenous water supplied by the water-spray that plays the more significant role.

Visual inspection of surfaces ablated without water spray exhibit charring and carbonization, which may be a good indicator that thermally damaging temperatures were generated on the surface of the tooth, as well as potentially the pulp. Morphological studies typically describe an ablated surface without spray as having a cracked and melted or “molten lava” like appearance with the deposition of non-apatite mineral fragments on the cavity preparation walls. The deposition of non-apatite debris (which would normally have been dispersed by a water spray) onto the preparation walls, as well the melted nature of the surface, may play a significant role in both the ability of a future restoration to be bonded, and the inability of the surface to be ablated further by the laser.
comparing these observations with those of an ablated surface cut in the presence of water spray, clean preparation walls with no smear layer, intact enamel prisms and dentin tubules, smoothly cut margins, and no evidence of charring or carbonization of the tissue are found.\textsuperscript{19,26}

Pulse rate and pulse energy also play important roles in the regulation of laser ablation rate. With respect to pulse energy or power (watts), a positive and linear correlation when compared to ablation rate is observed; as the pulse energy of the laser is increased, so too is the rate of ablation.\textsuperscript{7,8,11,46} Morphological observations studying changes on the target tissue surface with increasing pulse energy claim that as the power of the laser is increased, craters of greater diameter and depth are created.\textsuperscript{8,25,34} An increased likelihood of surfaces exhibiting a glassy or molten appearance has also been observed.\textsuperscript{11} Each of these morphological observations can be explained by the idea that a greater amount of energy is being delivered to the target tissue surface; therefore, a larger area of tissue can be vaporized or melted per pulse.

Pulse rate, also referred to as pulse per second (PPS), pulse repetition or frequency, is typically altered by changing the hertz (Hz) setting on the laser. Studies observing the effects of varying the pulse rate have found that increased
ablation is seen by increasing the number of pulses per second delivered to the target tissue.\textsuperscript{7,8} In fact, several studies that have compared the ablation rate of increased pulse energy versus increased pulse rate have found that pulse rate has a greater effect on the ablation rate than pulse energy.\textsuperscript{8} Lower pulse energy paired with increased pulse rate can be more effective during ablation than higher energy at lower pulse rates.\textsuperscript{37}

**Erbium Lasers and Intrapulpal Temperature**

In 1965, Zach and Cohen\textsuperscript{56} established the original threshold for irreversible damage to the pulp in their classical study which examined the effects of externally applied heat on the pulps of anterior and posterior teeth in rhesus monkeys. In the study, teeth were exposed to an external heat source until increases in temperature of 4ºF, 10ºF, 20ºF, and 30ºF were recorded. The teeth were then histologically examined at varying time intervals for signs of thermal trauma and irreparable pulpal necrosis. Fifteen percent of teeth with temperature increases of 10ºF developed irreversible pulpitis, compared to 60% of teeth heated to 20ºF, and almost no teeth that were heated below 10ºF. Based on these results, a threshold of 10ºF (5.5ºC) was set as the maximum temperature that the pulp can be heated to predictably minimize the chance of irreversible pulpal necrosis.
The pulpal response studies summarized in this section support the pulpal safety of erbium lasers used for the removal or alteration of dental hard tissues, if a thermal threshold of less than 5.5°C is observed. For the purpose of gathering a greater collection of data on the topic, studies relating to two types of erbium lasers, the Er,Cr:YSGG (2.78µm) and the Er:YAG (2.94µm), were considered. Both operate at nearly identical wavelengths, and both are absorbed by the same target substrates; water in the inorganic matrix, and –OH groups in hydroxyapatite of the inorganic matrix. Unfortunately, because laser settings can be customized into thousands of different combinations, and there are no standard settings for research, most of the studies reviewed use the same lasers with different settings.

In a 2003 study, Cavalcanti et al\textsuperscript{6} compared heat generated in the pulp by a conventional high-speed handpiece and a water-cooled Er:YAG laser (3.5W/10Hz) during Class V preparations on extracted bovine incisors. The water-cooled, high-speed handpiece showed an average temperature increase of 0.96°C, and the water-cooled laser showed an average increase of 2.69°C. Although the laser generated a higher temperature than the handpiece, both temperatures were found to be within the range of safety for the pulp. Klinic et al\textsuperscript{32} conducted a similar study in 2009 comparing an Er:YAG laser (700mJ/pulse), an Er,Cr:YSGG laser (5.5W), and a high-speed handpiece with different diamond
burs. This study also found that use of a traditional handpiece with diamond burs resulted in the lowest intrapulpal temperatures (0.46±1.14°C to -0.67±0.53°C), while the Er:YAG exhibited the highest temperatures (4.78±0.87 °C). The Er,Cr:YSGG laser resulted in a low, average temperature of 1.32±0.64°C, and all temperatures for all groups were below the 5.5°C benchmark.

Mollica et al\textsuperscript{39} and Firoozmand et al\textsuperscript{15} also observed intrapulpal temperature changes during Class V preparations on bovine incisors using a high-speed handpiece and an Er:YAG laser; however their results differed somewhat from the studies described above. Mollica et al reported a mean temperature rise of 1.10±0.56°C for the handpiece, and 0.84±0.55°C for the Er:YAG laser (250mJ/4Hz). Firoozmand et al showed 1.92±0.80°C for a low-torque handpiece, and 0.75±0.39°C for the Er:YAG laser (250mJ/4Hz). Unlike the studies conducted by Cavalcanti et al\textsuperscript{6} and Klinic et al\textsuperscript{32} these studies indicated that the laser resulted in a lower pulpal temperature increase. All of the temperatures reported were still below 5.5°C.

Krmeš et al\textsuperscript{32} conducted an Er:YAG study comparing heat generated in the pulp during Class V preparations with varying pulse energies and pulse rates. After completion of the study, they found that pulse energy appeared to be a stronger
influence on temperature increase in the pulp than pulse rate. The highest
temperature achieved in the study was 1.99±0.28ºC, indicating that the laser
ablated tissue with minimal increase in pulp temperature within a safe range.

In a different type of study, Burkes et al\textsuperscript{5} observed temperature increases in the
pulps of teeth ablated by an Er:YAG laser (56mJ, 60mJ, and 95mJ) with and
without water spray. Teeth irradiated without water spray exhibited intrapulpal
temperature increases of up to 27ºC while those irradiated with water spray
showed an average temperature increase of 4ºC. A few years later, Visuri et al\textsuperscript{52}
also made thermal measurements using an Er:YAG laser (10Hz, 360mJ/pulse)
with and without water spray on samples cut from human molars and premolars.
Without water spray, temperature increases that exceeded 15ºC were noted, while
with water spray, temperatures of less than 3ºC were observed. And finally, in the
same study by Cavalcanti et al\textsuperscript{6} described above, a pilot study on five teeth was
conducted before the main study to determine if testing the Er:YAG during cavity
preparation without water spray would provide valuable data. During cavity
preparation, temperatures of 40.86ºC were recorded with no ablation of the hard
tissue; therefore, the group was left out of the main study. Based upon these
results conducted using erbium lasers with and without water spray, it is found
that all tests conducted with water spray remained below 5.5°C, while those without water spray went significantly above.

In addition to cavity preparations, erbium lasers have also been indicated for use in caries prevention, tooth desensitization, and root canal sterilization. In 2008, de Freitas et al\textsuperscript{16} observed temperature increases in enamel blocks from third molars when an Er,Cr:YSGG laser programmed for caries prevention was used at 0.25W/20Hz, 0.5W/20Hz, and 0.75W/20Hz. The maximum increase in temperature observed during the study was 0.1°C; all samples were irradiated without the use of water spray.

It should be noted that when evaluating studies on endodontic treatment using erbium lasers, the temperature threshold of 5.5°C for the pulp no longer applies because it is the alveolar bone and periodontal ligament that are potentially affected, not the pulp. The alveolar bone is adversely affected by temperatures above 47°C (10°C increase from body temperature) for one minute;\textsuperscript{12} whereas the periodontal ligament has difficulty tolerating temperatures above 42-44°C (5-7°C increase from body temperature), depending on which study you prefer to use as a reference.\textsuperscript{22,35}
George et al\textsuperscript{20} conducted a study observing the thermal changes on the outside of an endodontically treated root treated by erbium lasers for sterilization. An Er:YAG (4W) and Er,Cr:YSGG (1.5W) laser were used to irradiate the canal of an amputated root, while temperature measurements were made on the external surface of the root. Temperature rises reached a maximum of approximately 2\(^\circ\)C, and a minimum of -2\(^\circ\)C. Ishizaki et al\textsuperscript{29} also conducted a study on thermal changes generated on the external surface of the tooth when the root canal was being irradiated for endodontic treatment. Maximum temperature rises for all wattages tested were found to be less than 8\(^\circ\)C. The results of both studies indicate that erbium lasers used for endodontic sterilization are most likely a safe procedure for both the alveolar bone and periodontal ligament.

**Current Literature on the Laser Removal of Ceramic Prostheses**

**A. Ceramic Orthodontic Brackets**

Very little research has been conducted on the use of lasers for the removal of ceramic prostheses. Studies on the subject appeared to have begun in the late 1980’s and early 1990’s with the idea that lasers could be used to remove newly introduced ceramic orthodontic brackets. Since that time, multiple PubMed searches have yielded only three articles relating to the laser removal of veneers, none of which deal with the potential thermal insult to the pulp. Studies on
ceramic orthodontic bracket removal are important because they offer
descriptions of the processes believed to be occurring which enable to laser to
debond the ceramic brackets from resin cement.

Studies observing the laser debonding of ceramic orthodontic brackets were
originally conducted in order to find a less destructive and more comfortable
method for the brackets to be removed. Up to the point of the studies, ceramic
brackets, which require significantly more force to remove than metal brackets,
were traditionally removed by forceps, or through direct heating of the resin with
an electrothermal instrument. Removal with forceps often resulted in enamel pull-
out, fracture of the bracket, and discomfort to the patient, while removal though
heating of the resin raised concerns over pulpal safety; therefore, a new method of
removing ceramic brackets was deemed necessary.

In 1992, both Strobl et al\textsuperscript{50} and Tocchio et al\textsuperscript{51} published papers comparing the
force needed to remove monocrystalline (single crystal/sapphire) and
polycrystalline orthodontic brackets with and without laser application. Strobl et
al\textsuperscript{50} observed that a significant reduction in force was necessary to remove
ceramic brackets that had been treated with either a CO\textsubscript{2} (10.6µm, 14W) or
Nd:YAG (1.06µm) laser. The explanation given was that the laser was
melting/softening the resin cement as the laser heated the composite beyond its “critical temperature,” which in this case was 150-200°C. It was also mentioned that monocrystalline brackets required less energy to achieve the same reduction in debonding force when compared to polycrystalline brackets. The reasoning behind this observation is that the monocrystalline brackets exhibit a more uniform distribution of crystals, and therefore allow a greater quantity of laser energy to pass through the material to interact with the resin-ceramic interface. Polycrystalline brackets are made of a variety crystals oriented in many directions which ultimately leads to greater absorption and diffusion of a larger part of the laser energy into the bracket itself which is then converted to heat.

Tocchio et al\textsuperscript{51} used three types of lasers at 248 (KrF), 308 (XeCl), and 1060\textmu m (Nd:YAG) wavelengths to debond mono and polycrystalline orthodontic brackets. When the lasers were applied to the brackets, it was observed that the single crystal alumina brackets “blew off” the tooth and were cool to the touch upon handling, while the polycrystalline alumina brackets “slid off” the tooth and were too hot to be handled. These observations indicated to the authors that two different types of debonding were occurring. The polycrystalline brackets, due to their crystalline arrangement, scattered incoming laser wavelengths, leading to the absorption of the laser energy and heating of the bracket followed by melting of
the resin cement beneath, which the authors described as *thermal softening*. The single crystal brackets, having a more uniform arrangement of crystals, allowed a larger portion of the laser energy to reach the resin cement. The laser energy then quickly vaporized a component of the cement (either water or residual monomer) and *thermal ablation* (as previously described in the mechanism of ablation section) or *photo ablation* occurred, which is why the brackets were cool. The resin became ablated before the bracket had time to heat. Although they did not measure intrapulpal temperatures during removal of the brackets, it was suggested in the study that the brackets removed by thermal softening may very well have generated enough heat to damage the pulp.

Using an Er:YAG laser at 4.2W, Oztoprak et al\textsuperscript{44} confirmed the conclusions of Tocchio et al\textsuperscript{51} and Strobl et al,\textsuperscript{50} finding that a significantly decreased force was necessary to debond laser-scanned, ceramic orthodontic brackets when compared to the force required to remove brackets that were not lased. In 1995, Rickabaugh et al\textsuperscript{R95} used a CO\textsubscript{2} laser (14W), and in 2011 Sarp and Gulsoy\textsuperscript{49} used a ytterbium fiber laser with the same results.

Hayakawa\textsuperscript{24} conducted a study similar to Tocchio et al\textsuperscript{51} and Strobl et al,\textsuperscript{50} using an Nd:YAG laser (1060nm) to remove both monocrystalline and polycrystalline
orthodontic brackets bonded to bovine teeth. In this study, however, intrapulpal temperature was also measured to assess potential harmful effects of laser debonding on the dental pulp. Upon laser application, carbonization was evident beneath both monocrystalline and polycrystalline brackets, indicating that the Nd:YAG laser was not heavily absorbed by the bracket, and that enough laser energy was able to interact with the resin to induce either thermal ablation or photoablation rather than thermal softening. The highest temperature generated in the pulp during the study was 5.1°C, indicating that the procedure may potentially be safe on human teeth. Similar intrapulpal temperature studies using 5.5°C as a threshold for pulpal damage conducted by Rickabaugh et al., Obata et al., Feldon et al., Sarp and Gulsoy, and Nalbantgil et al. on the laser removal of ceramic orthodontic brackets, also presented results that agree that the procedure is most likely safe for the pulp.

Azzeh and Feldon summarized current knowledge on orthodontic bracket debonding by lasers in a literature review. The conclusions arrived upon by the end of the review very closely mirror conclusions that may be gathered by reading the preceding paragraphs. The exact mechanism of laser debonding of orthodontic brackets needs further study, laser scanning of brackets can result in decreased
force necessary to debond, and if the correct parameters are used, brackets may be
debonded without causing significant thermal damage to the pulp.

B. Ceramic Veneers

As stated earlier, very few studies have actually been published on the laser
removal of porcelain veneers. PubMed searches yield only three articles on the
topic, and none that discuss thermal changes in the pulp. Although other studies
may claim to be the first article describing the laser removal of veneers, the actual
earliest article believed to have been published on the subject was a case study
written by Broome in 2005. In this study, an Er,Cr:YSGG laser at 4W, 20% H₂O,
40% Air, and 25Hz was used to remove eight feldspathic veneers in a patient for
esthetic purposes. Broome describes the mechanism by which the veneers were
removed as “reversing the light or dual-cure resin chemical reaction” or
“denaturing” the resin. It was stated that laser energy passes through the veneer to
interact with the ceramic-resin interface where it selectively excites water
molecules in the resin. Broome also went on to observe that feldspathic veneers
took between 5 and 30 seconds to remove, while pressed ceramics usually took
longer (between 20 seconds and 2 minutes), and that the dark areas of denatured
resin noted upon removal of the veneer were easily removed with a polishing cup.

Upon clinical observation, tooth structure beneath the veneer was said to be
unaffected by the laser, and only minimal corrections to the preparations were necessary for new veneers to be fabricated.

The next article to be published was by Oztoprak et al\textsuperscript{45} in 2011 on the effect of laser application duration on the debonding strength of veneers using an Er:YAG laser at 4.2W. The laser was scanned over the surface of 0.7mm thick, IPS Empress II, lithium disilicate discs bonded to extracted bovine teeth with Variolink II resin cement for 3, 6 and 9 seconds. Compared with the control group, to which the laser had not been applied, specimens tested after laser application required significantly less force to achieve debonding. The study also found an inverse relationship between laser exposure duration and debonding force; showing that as the laser exposure increased, the amount of force to debond the discs decreased. Oztoprak et al sites Tocchio et al\textsuperscript{51} when claiming that the mechanism of debonding for this study was thermal softening of the resin, but goes on to say that “it would be reasonable to expect there to be physical disruption of the luting composite rather than just thermal softening.”\textsuperscript{45} These conflicting statements indicate that the actual mechanism of debonding still requires further study.
The most recent study on the laser removal of porcelain veneers was conducted in 2011 by Morford et al, 40 months after the Oztoprak et al 45 study was published. Many facets of laser veneer removal were observed, including: the laser wavelength transmissibility/absorbance of e.max® (Ivoclar, Vivadent®, Schaan, Liechtenstein) and Empress® Esthetic (Ivoclar, Vivadent®, Schaan, Liechtenstein) ceramics and RelyX™ (3M ESPE, St. Paul, Minnesota) resin cement, laser energy transmission through e.max and Empress® Esthetic veneer materials, and the time required to debond a veneer after laser scanning.

According to Fourier transform infrared spectroscopy (FTIR) of the ceramic materials, both exhibited a strong absorption band in the 1,100nm range, which is most likely due to the silica content of the ceramic. Neither ceramic material exhibited strong absorption of bands in the 3,400 to 3,750µm range, indicating that H₂O and OH are not a major components, and that energy from lasers which operate in this range (i.e. an Er:YAG at 2940nm or Er,Cr:YSGG at 2,780nm) would not be strongly absorbed. RelyX resin cement, on the other hand, did show broad absorption in this range, indicating that H₂O and/or OH is available for laser energy absorption. Energy transmission readings for the Er:YAG (10Hz) laser used in the study showed that e.max transmitted between 26.5% and 43.7% of laser energy, whereas Empress Esthetic transmitted only 11.5% to 21% of the
energy. All veneers lased in the study were able to be removed, with an average debonding time of 113±76 seconds for Empress Esthetic restorations, and 100±42 seconds for e.max restorations.

**Statement of Purpose**

The purpose of this study is to determine if there is a significant difference in intrapulpal temperatures generated in extracted teeth with resin-bonded, ceramic veneers that have been scanned with Er,Cr:YSGG laser energy for the purpose of veneer debonding.

**Null Hypothesis**

There will be no significant difference between the intrapulpal temperatures generated in teeth with bonded veneers that have been scanned with Er,Cr:YSGG laser energy, and extracted teeth with veneers that have not been scanned with Er,Cr:YSGG laser energy.

**Specific Aims**

The specific aim of this study is to determine if laser removal of ceramic veneers is a safe and efficient dental procedure when proper laser settings are observed to protect the vitality of the dental pulp.
Significance

The laser-assisted removal of ceramic veneers can potentially increase patient comfort and reduce the amount of time and resources normally necessary to remove veneers with a high speed handpiece. Fewer resources utilized, less chair time for the patient and practitioner, and improved patient comfort can lead to reduced overhead and increased profits.
Chapter 2: Materials and Methods

Veneer Fabrication

Thirty, extracted, human premolar teeth, were stored in a 1% Chloramine-T solution. The teeth were prepared to receive IPS e.max Press HT (Ivoclar, Vivadent®, Schaan, Liechtenstein) lithium disilicate veneers according to the manufacturers’ recommended specifications. A high-speed handpiece, 0.5mm depth-cutting diamond bur and tapered diamond bur (Brasseler USA®, Savannah, GA) were used to achieve a veneer preparation depth of 0.7mm. Chamfered margins were prepared and extended 0.5mm beyond the mesio- and distobuccal line angles, as well as to ½ the thickness of the buccal cusp tip. The preparation margins did not extend onto the occlusal surface of the premolars, and all internal line angles were rounded.

Once veneer preparations on the extracted premolars were complete, a single veneer was waxed directly onto the surface of each tooth using a hard casting wax (Whip Mix®, Louisville, KY). Each veneer was waxed to a thickness of 0.7±0.1mm and this thickness was verified in three locations (the middle of the cervical third, the middle of the middle third, and the middle of the incisal third).
with the use of a wax caliper. Upon verification of veneer wax-up thickness, veneer patterns were prepared for investment.

Using an IPS® e.max Investment Ring System (200g) and IPS® Silicone Ring (200g) (Ivoclar, Vivadent®, Schaan, Liechtenstein), wax patterns were sprued according to the manufacturers’ recommended specifications and invested (four patterns per ring) using IPS® PressVEST Speed phosphate-bonded investment material (Ivoclar, Vivadent®, Schaan, Liechtenstein) (Figure 1). Invested patterns were allowed to set for 45 minutes before being placed directly into a burnout oven at 850ºC for 60 minutes. Once burnout of the patterns was complete, one IPS e.max® Press HT Ingot (Shade A2) (Ivoclar, Vivadent®, Schaan, Liechtenstein) was placed into the investment ring along with an IPS e.max® Alox Plunger (Ivoclar, Vivadent®, Schaan, Liechtenstein). The investment ring was then placed into a Programat® EP 5000/G2 (Ivoclar, Vivadent®, Schaan, Liechtenstein) press furnace and allowed to process utilizing the recommended, pre-programmed firing schedules.
Divestment was accomplished first by using glass beads at high-pressure to air-abrade the bulk of the phosphate-bonded investment ring. Once the pressed veneers were exposed, they were placed in a weak hydrochloric acid solution for ten minutes to loosen the reaction layer of investment still bonded to the veneer. After the acid bath, the reaction layer was removed from the veneer surfaces using low-pressure, glass bead abrasion. Each veneer was then sectioned from its sprue and the outer surface smoothed. At this point, the thickness of the veneer was verified to be 0.7±0.1mm thick in the same three locations as previously described to rule out casting distortion. To apply the final finish to the veneers, IPS e.max® Ceram Glaze Paste (Ivoclar, Vivadent®, Schaan, Liechtenstein) was applied to the
buccal surface of the restorations, and the restorations were placed back into press furnace on the pre-programmed glaze cycle (Figure 2).

![Figure 2: Ceramic Glaze Paste (left) and Completed Veneers (right)](image)

**Veneer Cementation**

Completed veneers were cemented to their respective extracted teeth according to the manufacturers’ recommended specifications. The internal surfaces of the veneers were first cleaned with 37% phosphoric acid for 15 seconds, then rinsed and dried, followed by the application of IPS® Ceramic Etching Gel (5% hydrofluoric acid) (Ivoclar, Vivadent®, Schaan, Liechtenstein) for 20 seconds to etch the surface of the ceramic (Figure 3). Monobond Plus (Ivoclar, Vivadent®, Schaan, Liechtenstein) universal primer was then applied to the internal surfaces of the veneers for 60 seconds and allowed to air dry (Figure 3). At this point, the surface of the preparation surface of the tooth was steam cleaned, dried, and
etched with 37% phosphoric acid for 30 seconds. The tooth was not allowed to
dry completely before ExciTE® F DSC (dual cure, single component) (Ivoclar,
Vivadent®, Schaan, Liechtenstein) (Figure 3) fluoride releasing adhesive was
applied to the prepared tooth surface, air-dispersed, and light-cured for 10 seconds
with a bluephase® G2 (Ivoclar, Vivadent®, Schaan, Liechtenstein) LED curing
light. Variolink® Veneer High Value +1 (Ivoclar, Vivadent®, Schaan,
Liechtenstein) resin-based cement was then applied to the internal surface of the
veneer, and the veneer was seated into place on the tooth. A two second cure with
the bluephase® G2 LED curing light was used to remove excess cement, the final
cure of the cement was accomplished afterwards with a 20 second light-cure.

Figure 3: Ceramic Etching Gel (left) Monobond Primer (right)
Experimental Setup

With the veneers bonded in place, the roots of the teeth were sectioned using a tapered diamond bur 2mm apical to the CEJ to expose the pulp chamber; any remaining pulpal tissue was removed from the chamber at this time (Figure 5). A hole the size of the remaining apical portion of the root was then cut into a 1.25 x 1.25in square of plastic, suck-down material (Keystone Industries, Myerstown, PA); the tooth was luted with sticky wax and super glue to the plastic sheet so that the coronal portion of the crown could be isolated from the access to pulp chamber. An OMEGA® K-Type thermocouple (OMEGA Engineering, Inc., Stamford, CT) was then tacked into place with sticky wax so that it was held firmly against the buccal wall of the pulp chamber directly beneath the cemented veneer. The thermocouple was held more firmly in place with the bulk application
of Blu-Mousse® (Parkell, Inc., Edgewood, NY) bite-registration material, being sure not to allow the Blu-Mousse® into the pulp chamber. Verification of the thermocouple placement was confirmed using a periapical radiograph (Figure 5).

![Radiographic Verification of Thermocouple Placement](image)

**Figure 5:** Radiographic Verification of Thermocouple Placement

Once the position of the thermocouple was verified, the entire apparatus was luted with sticky wax to a large, plastic cube fabricated to allow for complete isolation of pulp chamber and easy handling during lasing of the veneer. The thermocouple was connected to an HH506RA dual-input, high accuracy datalogger (OMEGA Engineering, Inc., Stamford, CT) which was in turn connected to a Sony Vaio® VGN-CS320J laptop computer (Sony Corporation, Tokyo, Japan) loaded with
HH506RA software (Figure 6). The software enabled data gathered from the thermocouple to be recorded in a Microsoft® Excel spreadsheet.

![Completed Experimental Setup](image)

**Figure 6: Completed Experimental Setup**

**Experimental Procedures**

Upon completion of the experimental setup, each veneer was randomly assigned to one of five different laser setting groups: 0W and 0Hz, 2.5W and 25Hz, 3.5W and 25Hz, 2.5W and 35Hz, or 3.5W and 35Hz (Figure 7); six teeth were assigned to each of the five groups. Water and air settings on the Waterlase MD® Er,Cr:YSGG laser (Biolase® Technology, Inc., Irvine California) remained at 30% and 70% respectively for all groups; the only settings changed during the course of the study were the power (watts) and pulses per second (Hz). An MZ6,
600µm, quartz tip (Biolase® Technology, Inc., Irvine California) was utilized on the Er,Cr:YSGG laser handpiece for all specimens.

![Waterlase MD® Er,Cr:YSGG Laser Interface](image)

**Figure 7:** Waterlase MD® Er,Cr:YSGG Laser Interface

To begin each trial, the mounted tooth with thermocouple in position was allowed to rest until the reading inside the pulp chamber stabilized at room temperature. At this point, the HH506RA computer software was programmed to take temperature readings at one-second intervals, and the stopwatch to record removal time was activated. The Er,Cr:YSGG laser tip was then positioned perpendicular to the surface of the veneer at a distance of 1-2mm in the mesial-cervical corner.
Laser “scanning” or “painting” of the veneers was completed in “passes,” and the first pass was completed in a vertical fashion from cervical to incisal, then progressed in a distal direction at 0.5mm increments until the most distal aspect of the restoration was reached. Upon completion of the initial pass, which typically averaged 30-40 seconds to complete, a discoid-cleoid hand instrument (American Eagle Instruments® Inc., Missoula, MT) was used to attempt to dislodge the restoration. If the restoration was could not be removed, a second pass was completed with the laser in a horizontal fashion beginning in the mesial-cervical corner of the veneer and progressing in a mesial to distal direction at 0.5mm increments until the incisal aspect of the veneer was reached. At this point, the discoid-cleoid hand instrument was again used to attempt to remove the veneer. This pattern continued until the veneer was removed, or a removal time of five minutes had elapsed. Upon removal of the veneer, or once the removal time reached five minutes, the stopwatch and HH506RA computer software was told to terminate.

Method of Statistical Analysis

For each dependent variable, the analysis was a 1-way analysis of variance (ANOVA) with a subsequent Ryan-Einot-Gabriel-Welsch Multiple Range (REGWQ) test to determine the significance of all pairwise comparisons. These
analyses were performed with the SAS program (SAS Institute Inc., Cary, NC, USA. Proprietary Software 9.2) using the ANOVA Procedure.
Chapter 3: Results

Initial Temperature

The mean initial starting temperature for each experimental group was maintained as close as possible to all other groups for the purpose of minimizing future statistical error. After averaging the initial starting temperatures (Table 5) for each of the six trials within a group, the mean initial starting temperatures for all groups were subjected to a one-way ANOVA with a subsequent REGWQ test to determine the significance of all pairwise comparisons using an $\alpha = 0.05$. Based upon the data presented in Table 1, it can be observed that all groups, except for group 3.5W/25Hz, did not differ significantly in mean initial staring temperature. Groups with the same alphabetical letter in the “significance” column of Table 1 did not differ significantly from one another.

<table>
<thead>
<tr>
<th>Group</th>
<th>Mean Initial Temp (°C)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>0W/0Hz</td>
<td>21.00</td>
<td>A</td>
</tr>
<tr>
<td>2.5W/25Hz</td>
<td>21.00</td>
<td>A</td>
</tr>
<tr>
<td>3.5W/25Hz</td>
<td>21.76</td>
<td>B</td>
</tr>
<tr>
<td>2.5W/35Hz</td>
<td>21.05</td>
<td>A</td>
</tr>
<tr>
<td>3.5W/35Hz</td>
<td>20.96</td>
<td>A</td>
</tr>
</tbody>
</table>

Table 1: Mean Initial Temperature and Significance by Group
As indicated in Figure 8 below, the initial starting temperature for all groups was approximately 21°C, the temperature of the room at the time that the trials took place. Not all trials were completed on the same day. The only group that differed significantly from the other groups was the 3.5W/25Hz group, which had a mean initial temperature of 21.7°C.

Figure 8: Mean Initial Temperature Chart
Maximum Deviation Temperature

The maximum deviation temperature, as defined by this study, is the single, recorded temperature for each trial that differs most from the initial starting temperature of that trial. As seen in Table 2, the mean maximum deviation temperature for each group was determined by averaging the six maximum deviation temperatures from the individual trials of each group (Table 5). The mean maximum deviation temperatures from each group were then subjected to a one-way ANOVA with a subsequent REGWQ test to determine the significance of all pairwise comparisons using an $\alpha = 0.05$. Results of the statistical analysis indicated that all groups, with the exception of group 2.5W/35Hz, differed significantly from one another. Group 2.5W/35Hz, was found to be statistically similar to groups 2.5W/25Hz and 3.5W/35Hz.

<table>
<thead>
<tr>
<th>Group</th>
<th>Mean Max Deviation Temp (°C)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>0W/0Hz</td>
<td>19.10</td>
<td>A</td>
</tr>
<tr>
<td>2.5W/25Hz</td>
<td>24.25</td>
<td>B</td>
</tr>
<tr>
<td>3.5W/25Hz</td>
<td>26.63</td>
<td>C</td>
</tr>
<tr>
<td>2.5W/35Hz</td>
<td>25.93</td>
<td>B,C</td>
</tr>
<tr>
<td>3.5W/35Hz</td>
<td>29.18</td>
<td>D</td>
</tr>
</tbody>
</table>

Table 2: Mean Maximum Deviation Temperature and Significance
Observations based on the data presented in Figure 9 indicated that when the average power (wattage) and/or pulse repetition rate (Hz) of the laser was increased, the maximum deviation temperature also increased. For example, in groups with the same average laser power, when the pulse repetition rate is increased, the maximum deviation temperature also increased. The same was true when comparing two groups with the same pulse repetition rate, but increasing average power.

**Figure 9:** Mean Maximum Deviation Temperature Chart
Maximum Deviation in Temperature

The maximum deviation in temperature is the temperature that represents the difference between the initial temperature, and the maximum deviation temperature for a given trial. As seen in Table 3, the mean maximum deviation in temperature for each group was determined by averaging the six maximum deviations in temperature from the individual trials of each group (Table 5). The mean maximum deviations in temperature from all groups were then subjected to a one-way ANOVA with a subsequent REGWQ test to determine the significance of all pairwise comparisons using an $\alpha = 0.05$. The results showed that all groups had maximum deviations in temperature that were significantly different than both the 0W/0Hz control group, and the 3.5W/35Hz group. The other groups, including: 2.5W/25Hz, 3.5W/25Hz and 2.5W/35Hz were statistically similar.

<table>
<thead>
<tr>
<th>Group</th>
<th>Mean Max Deviation in Temp (°C)</th>
<th>Significance</th>
</tr>
</thead>
<tbody>
<tr>
<td>0W/0Hz</td>
<td>-1.90</td>
<td>A</td>
</tr>
<tr>
<td>2.5W/25Hz</td>
<td>3.25</td>
<td>B</td>
</tr>
<tr>
<td>3.5W/25Hz</td>
<td>4.86</td>
<td>B</td>
</tr>
<tr>
<td>2.5W/35Hz</td>
<td>4.88</td>
<td>B</td>
</tr>
<tr>
<td>3.5W/35Hz</td>
<td>8.21</td>
<td>C</td>
</tr>
</tbody>
</table>

Table 3: Mean Maximum Deviation in Temperature and Significance
Observations based on the data presented in Figure 10 indicate that when the average power (wattage) and/or pulse repetition rate (Hz) of the laser was increased, the maximum deviation in temperature also increased. For example, at the same wattage, when the pulse repetition rate was increased, the maximum deviation in temperature also increased. The same was true when comparing two groups with the same pulse repetition rate, but increasing average power.

Figure 10: Mean Maximum Deviation in Temperature Chart
Time to Debond

The time to debond data presented in Table 4 represents the mean length of laser scanning time necessary to debond a veneer from a tooth at a given wattage and pulse repetition rate. A time limit of five minutes was set for the study; those veneers that were not debonded within the five minute time limit were given a value of 5+ minutes. As can be observed in both Table 4 and Figure 11, with the pulse repetition rate left the same, an increase in average wattage (i.e. 2.5W to 3.5W) decreased the amount of time necessary to remove a veneer. It was also shown, that at a given average power, an increase in pulse repetition rate (i.e. 25Hz to 35Hz) increased the time necessary to debond a veneer. It was not possible to debond veneers in the 0W/0Hz and 2.5W/35Hz groups within the five minute time limit, which is why they were given a mean debonding time of 5+ minutes.

<table>
<thead>
<tr>
<th>Group</th>
<th>Mean Time to Debond (min)</th>
<th>% Debonded</th>
</tr>
</thead>
<tbody>
<tr>
<td>0W/0Hz</td>
<td>5.00+</td>
<td>0%</td>
</tr>
<tr>
<td>2.5W/25Hz</td>
<td>2.43</td>
<td>100%</td>
</tr>
<tr>
<td>3.5W/25Hz</td>
<td>0.85</td>
<td>100%</td>
</tr>
<tr>
<td>2.5W/35Hz</td>
<td>5.00+</td>
<td>0%</td>
</tr>
<tr>
<td>3.5W/35Hz</td>
<td>1.82</td>
<td>100%</td>
</tr>
</tbody>
</table>

Table 4: Mean Time to Debond by Group
Figure 11: Mean Time to Debond
<table>
<thead>
<tr>
<th>Trial</th>
<th>Group</th>
<th>Initial Temp (°C)</th>
<th>Maximum Deviation Temp (°C)</th>
<th>Max Deviation in Temp (°C)</th>
<th>Time to Debond (min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>0.0W/0.0Hz</td>
<td>21.2</td>
<td>18.9</td>
<td>-2.3</td>
<td>5.00</td>
</tr>
<tr>
<td>2</td>
<td>0.0W/0.0Hz</td>
<td>21</td>
<td>19.6</td>
<td>-1.4</td>
<td>5.00</td>
</tr>
<tr>
<td>3</td>
<td>0.0W/0.0Hz</td>
<td>20.8</td>
<td>19.8</td>
<td>-1</td>
<td>5.00</td>
</tr>
<tr>
<td>4</td>
<td>0.0W/0.0Hz</td>
<td>21.2</td>
<td>19.2</td>
<td>-2</td>
<td>5.00</td>
</tr>
<tr>
<td>5</td>
<td>0.0W/0.0Hz</td>
<td>20.9</td>
<td>17.7</td>
<td>-3.2</td>
<td>5.00</td>
</tr>
<tr>
<td>6</td>
<td>0.0W/0.0Hz</td>
<td>20.9</td>
<td>19.4</td>
<td>-1.5</td>
<td>5.00</td>
</tr>
<tr>
<td>7</td>
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<td>20.4</td>
<td>23.8</td>
<td>3.4</td>
<td>3.01</td>
</tr>
<tr>
<td>8</td>
<td>2.5W/25Hz</td>
<td>21</td>
<td>24.2</td>
<td>3.2</td>
<td>3.15</td>
</tr>
<tr>
<td>9</td>
<td>2.5W/25Hz</td>
<td>20.7</td>
<td>24.6</td>
<td>3.9</td>
<td>0.83</td>
</tr>
<tr>
<td>10</td>
<td>2.5W/25Hz</td>
<td>21.8</td>
<td>25.6</td>
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<td>3.8</td>
<td>2.65</td>
</tr>
<tr>
<td>12</td>
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<td>21.7</td>
<td>23.1</td>
<td>1.4</td>
<td>1.43</td>
</tr>
<tr>
<td>13</td>
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<td>28.1</td>
<td>6.4</td>
<td>1.08</td>
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<td>0.66</td>
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<td>3.5W/25Hz</td>
<td>22.2</td>
<td>26.2</td>
<td>4</td>
<td>0.71</td>
</tr>
<tr>
<td>18</td>
<td>3.5W/25Hz</td>
<td>21.4</td>
<td>24.6</td>
<td>3.2</td>
<td>1.30</td>
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<td>21</td>
<td>22.8</td>
<td>1.8</td>
<td>5.00</td>
</tr>
<tr>
<td>20</td>
<td>2.5W/35Hz</td>
<td>21.3</td>
<td>26.3</td>
<td>5</td>
<td>5.00</td>
</tr>
<tr>
<td>21</td>
<td>2.5W/35Hz</td>
<td>20.8</td>
<td>24.8</td>
<td>4</td>
<td>5.00</td>
</tr>
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<td>22</td>
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<td>20.9</td>
<td>26.6</td>
<td>5.7</td>
<td>5.00</td>
</tr>
<tr>
<td>23</td>
<td>2.5W/35Hz</td>
<td>21.1</td>
<td>27</td>
<td>5.9</td>
<td>5.00</td>
</tr>
<tr>
<td>24</td>
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<td>21.2</td>
<td>28.1</td>
<td>6.9</td>
<td>5.00</td>
</tr>
<tr>
<td>25</td>
<td>3.5W/35Hz</td>
<td>21</td>
<td>28.7</td>
<td>7.7</td>
<td>1.98</td>
</tr>
<tr>
<td>26</td>
<td>3.5W/35Hz</td>
<td>21.2</td>
<td>30</td>
<td>8.8</td>
<td>1.56</td>
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<td>3.5W/35Hz</td>
<td>20.9</td>
<td>26</td>
<td>5.1</td>
<td>1.75</td>
</tr>
<tr>
<td>28</td>
<td>3.5W/35Hz</td>
<td>21</td>
<td>32</td>
<td>11</td>
<td>1.98</td>
</tr>
<tr>
<td>29</td>
<td>3.5W/35Hz</td>
<td>20.9</td>
<td>30.5</td>
<td>9.6</td>
<td>1.81</td>
</tr>
<tr>
<td>30</td>
<td>3.5W/35Hz</td>
<td>20.8</td>
<td>27.9</td>
<td>7.1</td>
<td>1.83</td>
</tr>
</tbody>
</table>

**Table 5:** Data Gathered from all Trials of Each Group
Chapter 4: Discussion

Pulse Repetition Rate (Hz)

Increasing the pulse repetition rate (Hz) setting on the laser led to an increase in debonding time at both average power (W) settings. At 2.5W/25Hz the veneers were debonded in a mean time of 2.43 minutes (2min 26sec), whereas at 2.5W/35Hz, the debonding time of the veneers exceeded the 5 minute time limit set for the study. A similar increase in debonding time was seen at the 3.5W average power setting. Veneers were easily debonded in an average time of 0.85 minutes (51sec) at 3.5W/25Hz, whereas increasing the pulse repetition rate to 35Hz increased the average time to debond to 1.82 minutes (1min 49sec). These observations may be explained by the following equation, which allows for the determination of pulse energy (the amount of energy being delivered per pulse) using the laser settings described above:

\[ J = \frac{W}{Hz} \]

**Figure 12:** Pulse Energy (J) = Average Power (W) / Pulse Repetition Rate (Hz)
If the laser settings for each group observed in the study are substituted in the equation from Figure 12, we find the following:

<table>
<thead>
<tr>
<th>Group</th>
<th>Pulse Energy (mJ)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0W/0Hz</td>
<td>0</td>
</tr>
<tr>
<td>2.5W/25Hz</td>
<td>100</td>
</tr>
<tr>
<td>3.5W/25Hz</td>
<td>140</td>
</tr>
<tr>
<td>2.5W/35Hz</td>
<td>71</td>
</tr>
<tr>
<td>3.5W/35Hz</td>
<td>100</td>
</tr>
</tbody>
</table>

**Table 6: Energy per Pulse by Group**

When comparing the pulse energies of the groups with the same average power setting, it was found that by increasing the pulse repetition rate, the amount of energy being delivered per pulse is actually decreased. This decrease in laser energy being delivered to the resin cement with each laser pulse may have contributed to the increase in debonding time (Table 4) observed in this study between groups of similar average power with differing pulse repetition rates.

Increasing the pulse repetition rate setting on the laser also led to an observed increase in measured intrapulpal temperatures between groups with similar average power settings. At 2.5W/25Hz, the mean change in intrapulpal
temperature was 3.25°C, which was less than the 4.88°C recorded at 2.5W/35Hz. The same results were also true when comparing intrapulpal temperatures recorded at 3.5W/25Hz and 3.5W/35Hz, which exhibited mean temperature increases of 4.87°C and 8.21°C respectively. The following diagram and accompanying equation, which allows for the calculation of the distance between peak pulse energies in time (t), may help to describe these observations:

Figure 13: Time between Peak Pulse Energy = 1/ Pulse Repetition Rate (Hz)
Using the equation from Figure 13, and the pulse repetition rates from each experimental group, the data presented in Table 7 is gathered.

<table>
<thead>
<tr>
<th>Group</th>
<th>Time between Peak Pulse Energy (sec)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0W/0Hz</td>
<td>0.00</td>
</tr>
<tr>
<td>2.5W/25Hz</td>
<td>0.04</td>
</tr>
<tr>
<td>3.5W/25Hz</td>
<td>0.04</td>
</tr>
<tr>
<td>2.5W/35Hz</td>
<td>0.028</td>
</tr>
<tr>
<td>3.5W/35Hz</td>
<td>0.028</td>
</tr>
</tbody>
</table>

**Table 7:** Time between Peak Pulse Energy by Group

Although less energy is being delivered per pulse at 35Hz (as illustrated in Table 6), when the time between pulses is observed (Table 7), it is found that at 35Hz there is less time between peak pulse energies than at 25Hz. This means that although less energy is being applied to the target tissue per pulse at an increased Hz, the tissue will actually have less time to cool (rest) between peak pulse energies. If the tissue has less time to cool between pulses, then this would potentially allow the opportunity for a greater increase in temperature despite less energy being applied per pulse at higher pulse repetition rates.
Average Power (W)

Based on the data obtained in Tables 2 and 3, it was observed that when comparing groups with the same pulse repetition rate, the group with the higher average power setting (W) resulted in an increased temperature rise intrapulpally. At both 25Hz and 35Hz, groups with the greater average power setting consistently yielded greater intrapulpal temperature measurements than their lower wattage counterparts. More importantly, all groups that were exposed to the laser had significantly higher maximum deviation temperatures and maximum deviation in temperature recordings than the control group (0W/0Hz), in which water and air were still being applied to the tooth, but the laser itself, was not. These observations can be explained using the data from Table 6 which shows the amount of energy being applied to the tooth per pulse. In groups with similar pulse repetition rates, it is seen that with increasing average power, an increased amount of energy is being delivered per pulse to the tooth. This increased energy being delivered to the tooth per pulse at a given pulse repetition rate is most likely the reason for the temperature increase noted at increased power settings.

It should also be noted that groups with higher average power settings were debonded more quickly than groups with lower average power settings, and similar pulse repetition rates. This could be explained by the idea that at higher
power levels, the resin cement beneath the veneer is being more effectively ablated than at lower power levels. Although the laser ablation threshold for Variolink Veneer is unknown, it could be assumed that at higher wattages, and therefore higher pulse energies, that the ablation threshold for the cement might be more easily achieved than at lower wattages. Evidence of this could be found in the observation that at higher wattages, larger areas of brown, charred/carbonized resin cement were noted on both the tooth and veneer surfaces.

**Thermal Damage to the Pulp**

Based strictly on the data presented in Table 3, it can seen that all laser setting groups, except 3.5W/35Hz, maintained mean maximum deviations in temperature below the 5.5°C benchmark for predictable pulpal safety set by Zach and Cohen in 1965.\(^56\) It should be noted, however, that all initial temperature settings in this study began at approximately 21°C, well below the average human body temperature of 37°C. Below is Figure 13, which graphs the maximum deviation temperature obtained from each experimental group against normal body temperature and normal body temperature plus the 5.5°C threshold increase.
Based on the data from Figure 13, it could be speculated that all of the experimental groups, including 3.5W/35Hz, would be well under the 5.5°C temperature threshold of an actual, vital dental pulp. Were this study to be redone, it would be recommended that the initial starting temperature for all trials be at 37°C for more clinically relatable results.
Ablation of the Resin Cement

As previously stated, upon removal of several veneer specimens, areas of what appeared to be charred/carbonized resin cement were often observed on both the surface of the prepared tooth and the internal surface of the veneer. This observation is consistent with the results of several studies\textsuperscript{11,19,26} which concluded that charring, carbonization, and melting/distortion of an affected tooth surface is more prominent when there is no water spray acting at the target site. Technically, as a veneer is being scanned with the laser for removal, the laser is acting at a site distant from the air-water spray. The air-water spray is actually acting directly on the external surface of the veneer, while a portion of the laser energy is propagating through the veneer to act at the resin-veneer interface. The cooling effect of the air-water spray must eventually find its way to the resin-veneer interface, but for a short time upon initial application, the laser is acting independent of any external coolant effect. It would not be surprising to find that intrapulpal temperature changes during laser veneer removal would be higher than intrapulpal temperature changes during laser cavity preparation at the same laser settings for this reason.

None of the veneers slid off the tooth upon removal, as former studies\textsuperscript{51} observing the laser removal of polycrystalline orthodontic brackets often noted; if this were
so, it might indicate that thermal softening of the resin cement, as defined by Tocchio et al\textsuperscript{51}, was the mechanism of removal. All veneers that were able to be removed “popped” off in one piece with minor force applied through a discoid-cleoid restorative instrument to the restoration margin. This observation, combined with the charring/carbonizing of the resin cement, most likely indicates that the veneers were removed by thermal ablation. The charring/carbonizing of the resin cement indicate that high temperatures were generated within the cement, most likely beyond the melting temperature of the resin, and into the ablative threshold (which occurs at a temperature higher than that of the melting temperature).

**Future Studies**

Based on the results and observations made during this study, the following potential future studies would prove beneficial to the understanding of the laser removal of ceramic restorations:

A. Some studies\textsuperscript{4,40} have noted that in various instances, fracture of restorations fabricated from certain ceramic materials occurred during laser removal (i.e. IPS Empress). An unpublished pilot study conducted by some of the authors of this study examining the laser removal of all-ceramic crowns also made this
observation. E.max restorations (Press or CAD), on the other hand, have not been noted to fracture in any study, including the current one. The question of whether laser debonded restorations have been weakened despite not fracturing has not been answered. An SEM study of un-fractured veneers after laser removal to check for microfractures that may have occurred within the ceramic is a possibility.

B. Another potential study would be to evaluate if bond strength is affected after the re-cementation of a laser debonded veneer. Does the laser have any effect on the internal surface of the veneer or tooth preparation surface that would reduce or increase the bond strength of a re-cemented veneer? Is there residual resin cement left on the surface of the tooth or veneer that would affect bond strength?

C. As previously stated, if this study were to be duplicated or redone, an excellent recommendation that would make the study more clinically relevant would be to start the trials at an initial temperature close to that of the vita pulp. Also, although human teeth of good quality are difficult to acquire, a greater sample size would also be recommended.
Chapter 5: Conclusions

Based on the results and observations gathered throughout this study, the following conclusions may be drawn:

A. The null hypothesis postulated at the beginning of the study was proven to be false. Based on the data from Tables 2 and 3, it is easily discerned that with the application of laser energy to a veneer bonded with resin cement to a prepared, extracted tooth, a significant increase in intrapulpal temperature was achieved when compared to the control group (0W/0Hz), in which no laser energy was used.

B. An increase of average laser power (W) and/or pulse repetition rate (Hz) resulted in an increase in intrapulpal temperatures generated within the extracted teeth.

C. An increase of average laser power (W) and/or decrease in pulse repetition rate (Hz) resulted in a decrease in the time necessary to debond the veneers.

D. The proposed mechanism of debonding for this study is thermal ablation.
E. Based on the results presented in Table 3, and the classic study by Zach and Cohen\textsuperscript{56} describing the threshold temperature of predictable pulpal necrosis as increases below 5.5°C, it can be inferred that all laser settings, except 3.5W/35Hz would be safe for use during veneer removal.

F. When observing the data from Table 5, it can be seen that no individual trial at 2.5W/25Hz resulted in a maximum deviation in temperature of greater than 5.5°C, and all veneers were able to be debonded in a mean time of 2 minutes 26 seconds. Although this group did not have the most efficient time for veneer debonding, it was the safest overall setting within the parameters of this study.
References


