# In Vitro Pulp Chamber Temperature Rise Created by Various Curing Lights During Orthodontic Bonding

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# Introduction

Concern over iatrogenic thermal insult to the pulp tissue is a long standing issue in dentistry. Several studies in the past have investigated the limits of intrapulpal temperature increase and the corresponding histologic response of the pulp tissue. The latest generation composite curing lights, called plasma arc curing lights, have inspired concern in this regard. The higher energy output of these lights produces a corresponding increase in the amount of heat generated. While these curing lights have been a boon for clinical efficiency, with cure times of one half to one fourth of conventional halogen curing lights, several investigators have questioned the possible thermal effect on the pulp tissue. More recent studies have examined intrapulpal temperature increase with regard to radiant energy from composite curing lights and the contribution of the exothermic reaction of composite polymerization. Prior research has focused on operative procedures, namely cavity preparations, and the proximity of the curing lights to the dental pulp. To date, no investigations have been undertaken to examine the possible increase in intrapulpal temperature caused by higher energy curing lights when used for orthodontic bonding purposes.

# Literature Review

Throughout the history of modern dental care, researchers and clinicians alike have been concerned that the operative procedures and techniques they are performing as well as the materials they are using may be causing iatrogenic harm to the vitality of the teeth. As each new piece of dental technology is introduced, studies have been undertaken to assess product safety and efficacy. One area of concern has been that of the susceptibility of pulp tissue to thermal changes; specifically, to an increase in intrapulpal temperature. Many operative procedures have been shown to cause an increase in intrapulpal temperature, such as cutting tooth structure with high-speed and slow-speed handpieces and polishing dental restorations.[3, 14, 21, 23, 32, 36] These procedures are subject to tremendous variability in terms of heat production depending on the operator's handling of and the type of equipment and material used such as the amount of air and water coolant, pressure applied, and so forth. The exothermic nature of composite polymerization reactions is subject to less operator variability and depends to a greater extent on the amount and type of material used.[2, 4, 13, 16, 17, 19, 20]

The frequently sited study by Zach and Cohen in 1965 has shown that increasing the intrapulpal temperature in excess of 42.5° C can result in irreversible pulp damage.[37] In their study, intrapulpal temperature rise was measured following the application of a heated soldering iron to the labial surface of lower incisors in a live Macaca rhesus monkey. Intrapulpal temperature increases ranged from 4 to 30° F. Following heat application, the teeth were histologically examined at two days, one week, two weeks, and fifty-six days. With an increase of less than 10° F, none of the teeth exhibited pulpal necrosis and were histologically indistinguishable from control teeth. However, of the teeth that measured at least a 10° F intrapulpal increase, 15% showed pulpal necrosis and all showed thermal stigmata and formation of irregular secondary dentin. An intrapulpal increase of greater than 20° F yielded pulpal necrosis in 60% of the teeth. Therefore, an intrapulpal temperature increase of 10° F or 5.5° C is often cited as the maximum permissible intrapulpal temperature rise prior to observance of a significant adverse tissue reaction.

Several studies have attempted to correlate the exact temperature at which externally applied heat will lead to histologic pulpal changes, in the absence of specific correlation to intrapulpal temperature rise. Zander and Lisanti applied a heated tip to cavities on the buccal aspect of the teeth of dogs at temperatures ranging from 125 to 600° F for ten seconds up to one minute.[38] The corresponding intrapulpal temperature increase was found to be 3 to 91° F. With the application of 300° F (149° C) for ten seconds edema and destruction of odontoblasts was seen, but over time this healed with irregular secondary dentin formation. At 600° F (316° C) for ten seconds, again no evidence of pulpal necrosis was seen. Postle et al exposed dog's teeth to a metal tip at temperatures of 215, 395, and 900° F (102, 201, 482° C) for twenty seconds.[25] No persistent damage was seen for the two lower temperatures; however, at the highest temperature some of the teeth exhibited necrosis. Nyborg and Brannstrom performed a similar experiment on human teeth in vivo under local anesthetic prior to extraction of the teeth for orthodontic treatment.[23] The teeth were subjected to 150° C for thirty seconds. The majority of the teeth showed localized changes such as cellular degeneration of odontoblasts or cell-rich zone and collagen formation, yet only a few teeth showed localized areas of necrosis.

Although the field of orthodontics is typically less invasive to tooth structure than other areas of dentistry, some orthodontic procedures have given rise to concerns regarding intrapulpal temperature increase. One such cause for concern has been that of the ETD or electrothermal debonding unit. Several studies involving the ETD have shown that, when used properly, this is a relatively safe method of debonding different types of orthodontic brackets and that the thermal effects on the pulp are minimal.[10, 27, 28, 29] Dovgan et al showed that use of the ETD produced histologic signs of inflammation and odontoblastic disruption but no pulpal necrosis.[10] Sheridan et al carried out a two part study on the ETD. Part one showed that in vitro, the average intrapulpal temperature rise was 3.2° C, or below the threshold established by Zach and Cohen.[28] Part two of the study was done in vivo and showed that aside from the pulpal response generally attributable to extraction of teeth, no adverse pulpal reaction to the ETD was seen histologically.[29] Rueggeberg and Lockwood showed that the temperature to debond using the ETD varied greatly depending on the type of bracket and the type of cement used.[27] The external temperature range required to debond the different types of brackets was from a low of 22° C for a powder and liquid adhesive system to 228° C for a two paste auto-cure composite system. The single mix or light-cure composite bonded brackets required a range of 61 to 167° C and a mean of 129° C. The external temperature applied to debond the brackets in some cases exceeded the temperature used in the Nyborg and Brannstrom experiment and the duration of heat application was longr at 38 to 68 seconds versus 30 seconds. The question may arise as to why pulpal necrosis was not seen in the Sheridan experiments when the EDT external temperature exceeds that which caused necrosis in some teeth in one experiment. An explanation lies in the fact that the intrapulpal temperature increase shown in the Sheridan experiments was below the threshold 5.5° C, most likely due to the shorter mean time reported of 7.8 seconds to debond the brackets. Also, as mentioned by Nyborg and Brannstrom in their literature review, even a moderate heat stimulus causes greater thermal insult when local anesthetic is used as was in their experiment. Presumably this is due to restricted blood flow from vasoconstricting components in the local anesthetic.

Worthy of brief mention is the use of high-speed and slow-speed handpieces and a variety of burs and attachments to remove orthodontic cement following debonding of orthodontic brackets and to polish the tooth structure thereafter. Here, as previously mentioned research indicates, when appropriate air or water coolants are used and excessive pressure is avoided, thermal insult to the pulp is averted. This is an aspect of clinical care that could theoretically lead to problems but is easily controlled by the operator.

Advances in dental technology in the modern arena are often driven by product marketing campaigns with seemingly less emphasis on product efficacy and safety. Recently, concern has been expressed regarding the new generation of high energy output composite curing lights. The concern lies in two primary areas. First, that the new lights generate a significantly greater amount of heat which may pose an iatrogenic threat to pulp vitality.[5, 11, 17, 18, 19, 26, 31, 34] Second, that the lights are not qualitatively curing composite on par with the previous generation of halogen curing lights.[9, 13, 15, 30, 33] Both of these concerns are worthy of consideration as these new curing light systems are gaining popularity in the orthodontic realm. Several studies have examined the use of high output curing lights in relation to temperature rise in operative dentistry applications; however, no study to date has examined the difference in intrapulpal temperature rise caused by the various curing light systems when used for orthodontic bonding purposes.

One of the most recent advents in composite curing light development has been that of high speed plasma arc lights. The plasma arc curing light system was designed with time efficiency in mind as dental offices are increasingly conscious of production per unit time. Manufacturers boast significantly reduced curing time at anywhere from 3 to 25 seconds for what is normally a 40 second recommended curing time. The plasma arc curing light consists of a high energy, high pressure ionized gas in the presence of an electrical current that creates a high temperature light source that is strong enough to increase the curing rate of composite resin.[11]

Several plasma arc light sources are commercially available.[9] The Apollo 95E (Dental Medical Diagnostic Systems, Woodland Hills, CA) has an output of 1370 mW/cm$^2$. Cure time settings can be 1, 2, and 3 seconds, or a step cure of ½ energy for 1.5 seconds and then full power for 4 seconds. The ADT 1000 PAC (American Dental Technologies, Corpus Christi, TX) has an output listed by the manufacturer as 240 to 750 mW/cm$^2$. The curing time is based on an auto-calibration unit consisting of a radiometer and a digital curing time display mounted on the front of the unit. Prior to each use, the output is measured by the radiometer and a cure time is set and registers on the display as to the amount of curing time required for a full cure. The curing time may range from 10 seconds to 25 seconds. The Kuring Light (Kreativ, Keltern, Germany) has an output of 1000 mW/cm$^2$ and has a setting for a standard cure time of 10 seconds, but may be adjusted to less. Despite the initial high cost of the units, the lamps in the plasma arc lights can work for 500 to 5000 hours, versus 50 hours for conventional halogen lights. Spectral output generally is narrow for plasma arc lights peaking at 470 nm.

Traditional halogen lamp curing lights have a spectral output of 360 to 500 nm which is generally wider than the 400 to 470 nm required by camphorquinone, the photoactivating element in visible-light cured composite.[9] One of the main variations between brands of curing light is the amount of energy output. This can be anywhere from 350 $mW/cm^2$ to as high as 650 to 800 $mW/cm^2$. Another variation in halogen lights is the two-step cure in which a low level output initiates the reaction, followed by a higher level output. This has been shown to be effective in increasing bond strength and other properties of the cured composite.

Several factors may influence the amount of curing time required by an individual curing light to achieve ideal results. Manufacturer recommended curing times vary from curing light to curing light and have been reported to be generally less than adequate when compared with qualitative analysis of the results.[24] This qualitative analysis includes flexural strength, modulus, and surface hardness.[15] Atmadja and Bryant found that a longer curing time results in greater hardness of the composite including time intervals that were longer than the manufacturer's recommendation.[2] They also reported that this increase in curing time did not increase the depth of the cure. Spectral distribution for various lights ranges from narrow (400-500 nm) to broad (300-800 nm) but must cover the range that activates the photoinitiator. Strang et al found that curing units vary in their spectral distribution and intensity for different models and even between units of the same make and model.[33] For each light tip, an intensity distribution may be mapped over the surface area of the light tip.[22] This intensity distribution may very greatly from uniform to highly irregular depending on the curing unit and the condition of the light tip. Light output versus distance adds another variable as between a distance of 2 and 10 mm the irradiance may decrease anywhere from 30 to 50%.[22] Light output as a function of time varies from -15% to +38% over a 60 second cure time.[22] Timers on curing lights have upon occasion been shown to be inaccurate by as much as 20%.[22] Other variables have been identified that affect the output of curing lights in general, such as equipment defects like a defective lamp or reflector, or the light tip may be contaminated or damaged.

Instrumentation is available for calibration of curing lights in order to estimate optimum curing times and eliminate the variable of intensity output. Digital radiometers register the amount of radiation. It has been shown, however, that the radiometer needs to be calibrated as well to ensure accurate readings. Hansen and Asmussen tested the reliability of three radiometers.[12] They found that some units that were measured as good by one radiometer were measured as poor by another. They also confirmed that there is occasionally a pronounced difference in the output measured for new curing lights of identical models from the same manufacturer.

Several studies have been undertaken to determine the amount of temperature rise created by various curing light sources. Smail et al found that a mean temperature increase of 12.2° C was produced in vitro when measuring the curing light temperature alone, and a mean of 5.5° C when measuring the temperature while curing a posterior composite material.[31] Hansen and Asmussen measured the temperature rise at the surface interface of the composite being cured and found an increase that ranged from 3.6 to 29.2° C, while 3.2 mm below the surface a temperature increase of 1.5 to 12.3° C was identified.[13] Powell et al found that conventional curing lights generate more intrapulpal heat (means of 7.6

to 14.7° F) than argon laser lights (2.7 to 3.0° F).[26] Thompson et al investigated the effect of various thicknesses of dentin using a standard curing light and found that a thickness of 2-2.5 mm gave a mean intrapulpal increase of <1.5° C while a thickness of <2 mm gave an increase of <2° C.[34] Bodkin and Share measured the ultimate temperature output at 120 seconds of ten curing lights at the light tip and found an output range of 90 to 130° F.[5] Two of the means they site as significant in enough to cause insult to the pulp tissue. Bennett et al compared intrapulpal temperature increase between auto-polymerizing and visible light-cured composites and found that the increase ranged from 0.41° C for the chemically cured resin to 3.81° C for the light-cured resin.[4] Losche and Roulet reported that irradiation time was a significant factor for temperature increase and that the steepest increase was during the first 20 seconds.[18] Lloyd et al has sited that there are two factors involved in temperature rise generated during the curing of composites. One is the exothermic reaction of the composite material itself, and second is the light system used to initiate the cure.[17] The composite alone contributed at most 1.5 to 3.4° C while the curing lights in combination with the composite were shown to be capable of generating a mean temperature increase of 21.4° C. Masutani et al reported that temperature rise when light-curing various composites with different sources ranged from 8.3 to 22° C.[19] Upon applying the light source to the already cured resin, they were able to subtract out the exothermic contribution of the composite which they found ranged from 3.1 to 8.9° C. McCabe studied the temperature rise of light-cured composite and reported a temperature increase of 20 to 40° C as measured by differential thermal analysis.[20] Hannig and Bott found that the mean intrapulpal temperature increase in a Class II cavity preparation with a 1 mm layer of dentin while curing a 2 mm depth of composite was more than 6° C for the ADT 1000 PAC and the Optilux 500 lights.

The current study will examine the intrapulpal temperature rise created by three different curing light systems when used for orthodontic bonding. A plasma arch curing light, a standard halogen light, and a modified fiber-optic light tip will be used. Of particular interest will be comparisons between the three lights as to whether one generates significantly greater heat than the others. Also of interest is a comparison of the temperature rise obtained to the physiologic threshold of 5.5° C established by Zach and Cohen above which pulpal necrosis may result. This will allow conclusions to be drawn as to the clinical significance of the temperature increase. Finally, the study will allow an evaluation as to whether the exothermic composite polymerization reaction has a significant contribution to the rise in intrapulpal temperature during orthodontic bonding. Trials will include a baseline reading at the tip of each light, intrapulpal temperature rise with the light tip on the clean tooth surface, and intrapulpal temperature rise when bonding orthodontic brackets.

# Materials and Methods

Ten extracted human lower incisors were selected for the experiment. The teeth were in good condition being caries and restoration free with clinical crowns and root structures intact. The extracted teeth had been stored in a ten percent Formalin solution and were transferred to a dilute sodium hypochlorite and water solution several hours prior to use. Immediately prior to use, each tooth was rinsed well in tap water and dried. Many more teeth were selected than were used in the experiment as when access was created to the pulp chamber from the apical aspect, several of the teeth had a calcified canal such that the thermal probe was not able to reach the most coronal aspect of the pulp chamber.

An Omega Engineering, Inc. 871A digital thermometer (Fig. 1) was used to

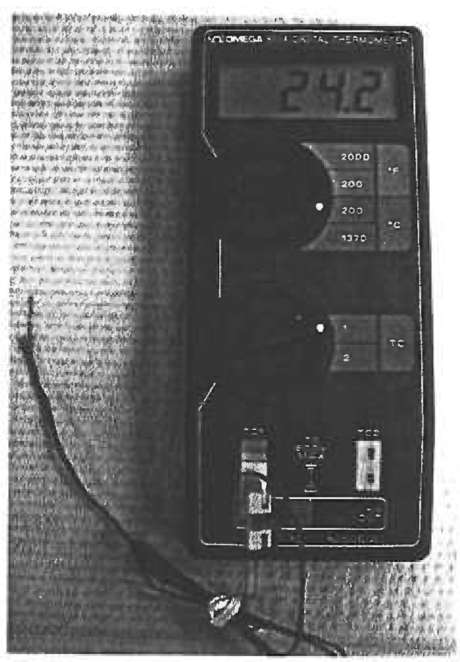


Fig. 1

take temperature readings. The thermometer was set to read on a Celsius scale measuring to the nearest tenth of a degree. The thermal probe was insulated except for the last few millimeters which were inserted into the tooth. Access to the pulp chamber was gained from the apical aspect of the root using a high speed handpiece and a 557 bur. An opening large enough to gain access to the pulp canal was created and located on the lingual root surface such that several millimeters of pulp canal were interposed between the access and the pulp chamber. All remaining pulp residue was extricated from the canal and pulp chamber using a Hedstrom endodontic file. The pulp canal and chamber were then dried of any remaining moisture using an air-water syringe. The thermal probe was then inserted such that it made contact with the most coronal aspect of the pulp chamber. The position of the probe was verified radiographically. The diametrical dimension of the pulp canal and chamber in the lower incisors was ideal for the size of probe used as the probe could be inserted and held firmly in place via friction without the aid of adhesives or other compounds that may act

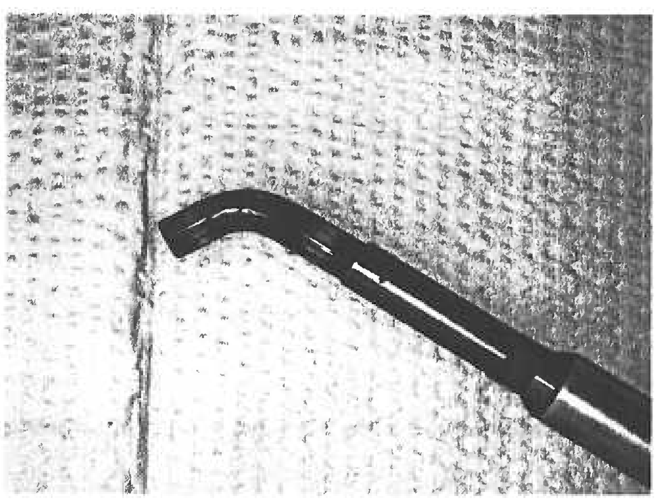

Fig. 2

as heat sinks or insulators.

The first curing light used was the ADT 1000 PAC Plasma Arc Curing System (American Dental Technologies, Corpus Christi, Texas) (Fig. 2). This light has a 430 to 500 nm optical output bandwidth with a curing power output range of 240 to 750 $mW/cm^2$ and a fiber-optic tip 6.5 mm in diameter according to the manufacturer. The ADT 1000 PAC has an auto-calibration feature in which a built-in radiometer takes a reading from the light tip and adjusts the full cure time based on the measured output. The range of this full cure time is from ten seconds to twenty five seconds. This feature is necessary as the light tips in this system are autoclavable and are changed after each use, each tip yielding a different output based on a number of factors including time in service and cleanliness and condition of the tip. The manufacturer recommends that for composites with recommended curing times of forty seconds, ten seconds is adequate with the ADT 1000 PAC. For recommended curing times of twenty seconds, the ADT 1000 PAC requires only five seconds of cure time. The light tip used for this experiment yielded a full cure time of twenty seconds upon auto-calibration (Fig. 3), thus the time increments for the experiment began at ten seconds rather than five.

The second curing system used was the Ortholux XT Visible Light Curing Unit (3M Unitek Corporation, Monrovia, California) (Fig. 4). This unit is a standerd halogen light curing unit. For the Ortholux XT light, the manufacturer

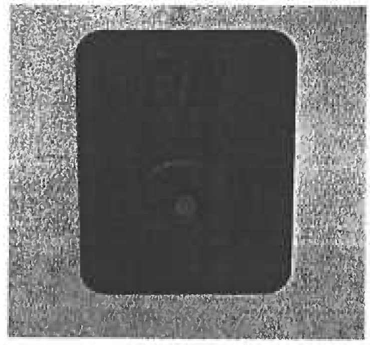

Fig. 3

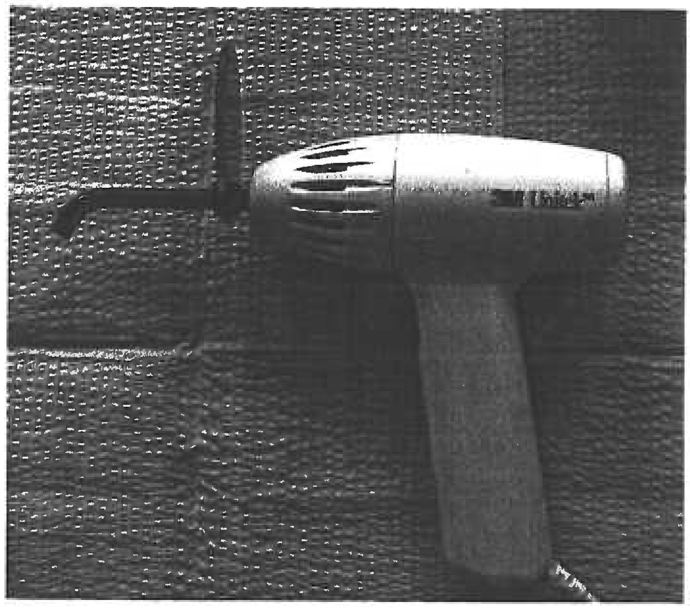

Fig. 4

recommends a curing time of ten seconds interproximally on both sides of the bracket for a total of twenty seconds.

The final curing light system used was the Reliance Power Slot (Reliance Orthodontic Products, Inc., Itasca, Illinois) (Fig. 5). The Power Slot is a fiber-optic tip that is compatible with most standard halogen curing lights. According to the manufacturer, the patented "tapered optics" increases the output of the curing

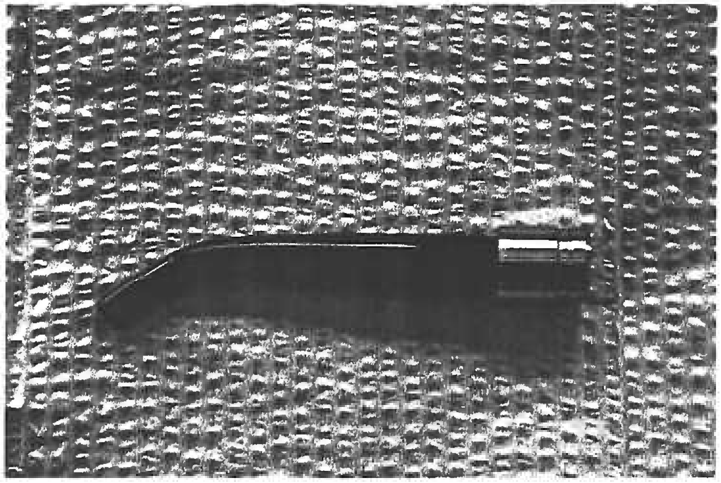

Fig. 5

light two and a half times allowing curing times of roughly one half that of standard halogen fiber-optic light tips. This curing light has a 420 to 500 nm optical output bandwidth and a 7 mm fiber-optic tip.

For the experiment, a baseline reading was obtained by measuring temperature increase at the end of the fiber-optic tip of each of the three different curing light systems. The curing time intervals used were based on the manufacturer recommended curing times and were 10, 20, and 40 seconds for the ADT 1000 PAC. For the Reliance Power Slot, intervals of 5, 10, and 20 seconds were used. And finally, for the Ortholux XT, intervals of 10, 20, and 40 seconds were used. Five trials were carried out for each curing light at each time interval for a total of forty-five trials.

After a baseline reading for each light was obtained, a baseline reading for each of the ten teeth was obtained. This was accomplished by inserting the thermal probe into the pulp chamber and then verifying its position radiographically (Fig. 6). The temperature increase within the pulp chamber during curing was then measured by placing the fiber-optic tip on the clean tooth surface centered in the middle third of the labial aspect of the crown and curing for each of the three time intervals for each of the three curing light systems for a total of ninety trials.

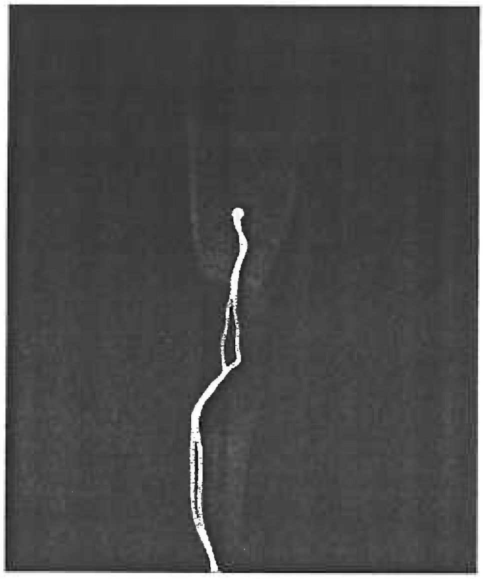

Fig. 6

Next, the tooth surface was prepared for bonding using Ultradent's Ultra-etch 35% phosphoric acid gel. The tooth was rinsed and dried and a thin coat of 3M Unitek's Transbond XT light cure adhesive primer was applied which was cured for ten seconds (Fig. 7). A 3M Unitek Victory Series APC incisor bracket was then placed. The APC series brackets are pre-coated with 3M Unitek's Transbond XT light cure adhesive paste. The excess adhesive paste was removed,

Fig. 7

and the bracket position was standardized at 4.5 mm from the incisal edge using a Boone bracket gauge. When the temperature reading from the probe had returned to room temperature and stabilized for several seconds, the bracket was cured in place from the mesial aspect. A bracket was cured and the intrapulpal temperature was measured for each of the three time intervals for each of the three light curing systems for a total of ninety trials. After each trial, the bracket was debonded and the tooth surface was cleaned using a high-speed handpiece with air coolant only and a Brasseler H375RF 016 carbide composite finishing bur using great care not to alter the enamel surface preparatory to the next trial.

# Results

The mean temperature increase and standard deviation was calculated for all trials (Appendix i, ii, iii). A one-way ANOVA test was performed to compare the differences in temperature rise and curing time increments between and within groups; a significant difference at the $p < .05$ level was found. Next, three independent statistical tests (Fisher's PLSD, Scheffe, and Bonferroni/ Dunn) were performed to determine the level of significance between the three different curing lights for each of the curing time increments within the baseline, tooth only, and bracket bonding groups of trials. Results identified which of the time increments yielded temperature increases that varied significantly between the three light sources. For the baseline curing light only trials (Appendix iv), temperature increase varied significantly between the three time increments for each of the lights. The only exception was with the Ortholux light for the comparison of the 10 to 20 second time increment.

For the tooth only trials (Appendix v), the first time increment yielded a significant temperature increase in only one category, which was the comparison of the ADT 1000 PAC to the Power Slot. The second time increment yielded no significance in the amount of temperature increase between the curing lights. The third time increment yielded a significantly greater temperature increase between the ADT 1000 PAC and the Power Slot, and between the Ortholux and the Power Slot, but not between the ADT 1000 PAC and the Ortholux.

For the bracket bonding trials (Appendix vi), comparison of the first and second curing time increments between the three lights did not yield a significant temperature increase. The third time increment yielded a significant temperature increase for the comparison of the ADT 1000 PAC and the Power Slot and the

Ortholux and the Power Slot, but not between the ADT 1000 PAC and the Ortholux.

For the baseline trials measuring temperature increase at the curing light tip alone, as curing time increased, temperature increased consistently in a linear fashion for all three lights for all curing time increments. This was also true for the tooth only and the bracket bonding trials. The average temperature increase for the baseline light only and tooth only trials was greatest for the ADT 1000 PAC light, followed by the 3M Unitek Ortholux light. For the bracket bonding trials, this was reversed with the 3M Unitek light yielding the greatest temperature increases consistently. The Power Slot light tip yielded the smallest temperature increase in all trials.

The maximum intrapulpal temperature increase obtained for all trials was 8.4° C. Of the baseline trials measuring temperature increase at the curing light tip alone, 28 of the 45 or 62% of the trials yielded temperature increases above 5.5° C. In trials where the curing light was placed on the clean tooth surface, 4 out of the 90 or 4.4% of the trials yielded intrapulpal temperature increases beyond 5.5° C. Of the trials in which a bracket was bonded to the tooth, 9 of the 90 or 10% of the bracket bonding trials caused an intrapulpal temperature increase above the threshold 5.5° C.

Certain of the teeth yielded consistently higher or consistently lower temperature increases across all trials. Tooth #10 yielded a total intrapulpal temperature increase for all trials of only 1.8° C for the tooth only trials, and 1.4°

C for the bracket bonding trials. Tooth #7 yielded a total intrapulpal temperature increase of 24.9° C for the tooth only trials, and 43.6° C for the bracket bonding trials.

# Discussion

A direct linear relationship was discovered between the amount of time the curing light was activated and the average amount of heat generated. As the curing time increment increased, the intrapulpal temperature also consistently increased. Consideration for this information has practical applications for the manufacture of curing lights and for the clinician. The ideal curing light would have an energy or irradiant output that would allow rapid curing times but would not exhibit a corresponding increase in the amount of heat generated. In the baseline light only trials, the heat generated at the light tip of the ADT 1000 PAC was consistently greater than that of the other two lights. To consider this from a different viewpoint, the same supposed curing result from the ADT 1000 PAC gained in a shorter time increment produced a disproportionately larger amount of heat than the standard curing light. Care should be given by the clinician that curing times in excess of the manufacturer recommended curing times should be avoided. One promising new candidate is that of the LED curing lights that are currently in development. Unlike conventional curing lights, these lights do not generate heat.[9] The argon laser has also been tested for orthodontic bracket bonding purposes with promising results. In vitro intrapulpal temperature rise has been measured to be significantly less (3° F or less) than that of conventional halogen curing lights.[26]

The average temperature rise measurement for the trials in which a bracket was bonded was in all cases higher than the trials using the tooth only as a baseline. For example, the average temperature increase for the ADT 1000 PAC for the 40 second tooth only trial was 3.12° C. The average temperature rise for the bracket bonding trial at 40 seconds was 4.01° C. Several explanations may be considered. First, as has been related by numerous researchers, the reaction produced in light-curing composite is exothermic in nature.[4, 5, 16, 17, 19] Thus the composite polymerization reaction itself may have contributed significantly to the temperature rise from the light source. Indeed, some of the variability between trials for each tooth may be attributable to the fact that an exactly consistent amount of composite was probably not present. Previous studies indicate that the amount of exothermic heat generated by the composite polymerization reaction is directly proportional to the amount of composite used.[6, 20] Second, the coefficient of thermal conductivity of the metallic bracket may have served to store and conduct the heat from the light to a greater degree than the enamel and dentin only in the tooth only trials. To eliminate this as a possibility, trials would need to be conducted that cure a uniform amount of composite only on the tooth surface, followed by curing a bracket with the same amount of composite on the tooth surface.

The differences in the average temperature increase between the ADT 1000 PAC and the 3M Unitek Ortholux light were slight but consistent. This is consistent with the data obtained by Hannig and Bott in their 1999 study using the same two curing lights.[11] It is interesting to note that the Ortholux light yielded higher average intrapulpal temperature increase than the ADT 1000 PAC for the bracket bonding trials, as this was not the case in the baseline and tooth only trials. Harrington in a 1996 study that tested the Optilux light rated it as one of the highest output halogen lights available. Certainly there are several other halogen curing lights on the market that could have been used that would have had a significantly lower power output and thus presumably lower relative temperature increases. According to Harrington, the ICI Luxor (Macclesfield, U.K.) has a predicted radiation time of 50% greater than that of the Optilux light. Also worthy of mention is the fact that several researchers have reported on the qualitative nature of the composite cured with plasma arc lights versus standard curing lights. It has been shown that flexural strength, modulus of elasticity, and surface hardness fail to reach adequate levels in some cases when plasma arc lights are used with manufacturer recommended curing times of short duration.[15, 30] As was mentioned earlier, studies have shown that manufacturers tend to report curing times of inadequate duration. If a less than ideal cure were achieved with the ADT 1000 PAC in the bracket bonding trials for this study, it would stand to reason that perhaps less of an exothermic reaction was obtained and thus the slight decrease in temperature rise relative to the Ortholux light for the bracket bonding trials.

Several of the individual trials caused intrapulpal temperature increases above the safety threshold established by Zach and Cohen of 5.5° C. The question may be considered as to why pulpal necrosis is not more frequently attributable to curing light induced intrapulpal temperature increase. Several explanations may be considered. First is the possibility that pulpal blood flow may aid in dispersing the heat and in preventing the significant build-up of heat in any one area of the pulp sufficient to cause significant damage.[21] A previous study has reported that in rat teeth under local anesthetic pulpal response to moderate temperature increase is significantly greater than for teeth not under the influence of local anesthetic.[23] This is presumably due to the effect of epinephrine in the local anesthetic causing reduced blood flow. Another explanation is that the surrounding tissue acts to draw heat from the tooth and dissipate the heat over a larger area. Consideration should be given to the difficulty in recognizing clinically the effect that an increase in intrapulpal temperature may have had on the pulp tissue. Even the teeth in the in vivo research that were subjected to 150° C for 30 seconds and then left in the mouth for up to two months prior to extraction gave no clinical signs in the form of symptoms of underlying histologic changes. From a histologic view-point, nearly all teeth subjected to even moderate intrapulpal temperature increases had persistent evidence of thermal insult in the form of disrupted cell-free zones, odontoblast layers, and ultimately the formation of irregular secondary dentin.[37] Another explanation may be that although the intrapulpal increase in temperature may cross the 5.5 C° threshold, this increase in temperature is transient enough that significant tissue damage is not incurred. In the previously sited in vitro and in vivo experiments, a much higher temperature was applied to the external surface of the tooth than is created by the curing light tips. Intuitively, a much hotter heat source would cause a greater temperature increase in the tooth structure immediately in contact with it such that the heat generated would take longer to dissipate and may in the meantime yield prolonged increases in intrapulpal temperature. Finally, in operative procedures the ultimate demise of the pulp following a restorative procedure may be easily attributable to mechanical insult to the tooth. For orthodontic bonding purposes, necrosis of the pulp in a tooth in the absence of any other compromising factors may rarely if ever be attributed by the clinician to increased pulpal temperature from a curing light.

Several etiologic factors may be involved when pulp tissue becomes necrotic. Fluctuations in pulpal blood flow may occur. The quality and quantity of pulp tissue may vary greatly depending on the age of the patient and parafunctional habits. The presence, size and condition of restorations on the teeth may contribute. Local bacteria populations and periodontal involvement of the teeth may contribute. Also, accounting for the significant variation between the teeth in terms of the total amount of temperature change experienced in the pulp chamber is the amount and character of the interposed dentin and enamel. The thermal conductivity of the overlying tooth structure may vary considerably depending on the relative thicknesses of enamel and dentin, as dentin has a thermal diffusivity of over 250% greater than that of enamel.[7] It is not inconceivable that thermal insult from which a normal pulp would completely recover may cause the ultimate necrosis of a compromised pulp. It is likely more often a combination of factors that lead to pulpal necrosis rather than strictly a single factor.

# Conclusions

1. A statistically significant difference was found between the ADT 1000 PAC, the 3M Unitek Ortholux, and the Reliance Power Slot curing lights for the amount of heat generated in the baseline trials. A significant difference in temperature rise in the tooth only and bracket bonding trials was only obtained at the longest curing increment and only between the Power Slot and the other two lights.
2. The amount of heat generated was linearly related to the amount of curing time; as the curing time increased, the amount of heat generated increased.
3. The amount of heat generated was greatest for the ADT 1000 PAC followed by the Ortholux and finally the Power Slot. This was true for the baseline and tooth only trials, but the Ortholux generated more heat in the bracket bonding trials than did the ADT 1000 PAC. This may have been affected by the exothermic nature of the composite polymerization reaction being qualitatively different for the ADT 1000 PAC versus the Ortholux.

4. In all cases, the temperature increase for the bracket bonding trials was greater than for the tooth only trials. The exothermic contribution of the curing composite was most likely responsible for this increase.
5. Several of the trials generated an intrapulpal temperature increase that was above the threshold reported by Zach and Cohen of 5.5° C above which 15% of the teeth examined were necrotic. Whether this is clinically relevant could be ascertained through in vivo bracket bonding trials and histologic evaluation of the pulps.
6. Care should be taken when using plasma arc lights that manufacturer's recommendations are supplanted with radiometric data that confirm the light's efficacy. The use of a calibrated radiometer is recommended. Curing times are generally longer based on this information than the manufacturer's recommendations as was the case in this study. As a result of this increase in curing time and the fact that the temperatures generated are greater than for conventional lights, clinicians may be legitimately concerned about iatrogenic pulpal thermal insult that could potentially be caused by these lights.

# Appendix i

## BASELINE-CURING LIGHT ONLY

| | ADT | | | POWER SLOT | | | OPTILUX | | |
|---|---|---|---|---|---|---|---|---|---|
| | TEMPERATURE INCREASE C | | | TEMPERATURE INCREASE C | | | TEMPERATURE INCREASE C | | |
| TIME (sec) | 10 | 20 | 40 | 5 | 10 | 20 | 10 | 20 | 40 |
| Trial 1 | 5.7 | 6.7 | 7.6 | 2.2 | 2.9 | 3.4 | 5.1 | 5.7 | 7.8 |
| Trial 2 | 5.6 | 7.1 | 7.9 | 2.1 | 2.9 | 3.3 | 5.5 | 5.6 | 7.4 |
| Trial 3 | 5.6 | 6.8 | 7.7 | 2.3 | 3.1 | 3.4 | 5.1 | 6.3 | 7.6 |
| Trial 4 | 6.2 | 6.6 | 7.8 | 2.1 | 2.7 | 3.3 | 5.6 | 6.7 | 6.1 |
| Trial 5 | 5.7 | 6.8 | 7.8 | 2.2 | 2.7 | 3.1 | 5.6 | 6.1 | 7.1 |
| **AVG** | 5.76 | 6.80 | 7.76 | 2.18 | 2.86 | 3.30 | 5.38 | 6.08 | 7.20 |
| **STDDEV** | 0.25 | 0.19 | 0.11 | 0.08 | 0.17 | 0.12 | 0.26 | 0.45 | 0.67 |

# Appendix ii

## CURING LIGHT ON TOOTH ONLY

| | TOOTH | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | TIME (sec) | TEMPERATURE INCREASE C | | | | | | | | | | AVG | STDDEV |
| ADT PAC | 10 | 0.6 | 1 | 0.3 | 0.5 | 0.4 | 1.3 | 1.6 | 2.8 | 0.8 | 0.2 | 0.95 | 0.79 |
| | 20 | 0.9 | 1.7 | 0.7 | 0.7 | 1.7 | 2.1 | 1.8 | 3.1 | 1.5 | 0.2 | 1.44 | 0.84 |
| | 40 | 1.5 | 5 | 1.5 | 0.9 | 2 | 4.8 | 6.3 | 6.9 | 2 | 0.3 | 3.12 | 2.39 |
| POWER SLOT | 5 | 1.1 | 0.6 | 0.1 | 0.7 | 0.3 | 0.2 | 0.8 | 2.5 | 0.2 | 0 | 0.65 | 0.74 |
| | 10 | 1.7 | 0.9 | 0.3 | 0.9 | 0.6 | 0.7 | 1.7 | 3.9 | 0.6 | 0.1 | 1.14 | 1.10 |
| | 20 | 0.4 | 0.6 | 0.6 | 1.5 | 1.2 | 1.4 | 2.7 | 4.9 | 0.8 | 0.2 | 1.43 | 1.42 |
| OPTILUX | 10 | 1.5 | 0.3 | 0.1 | 0.5 | 0.4 | 0.6 | 1.6 | 3 | 0.5 | 0.2 | 0.87 | 0.91 |
| | 20 | 1 | 0.6 | 0.5 | 0.9 | 1.2 | 0.8 | 2.2 | 4.4 | 0.9 | 0.2 | 1.27 | 1.22 |
| | 40 | 1.6 | 2.8 | 1.2 | 1.2 | 5.2 | 1.7 | 6.2 | 7.9 | 1.4 | 0.4 | 2.96 | 2.55 |
| | **TOTAL** | 10.30 | 13.50 | 5.30 | 7.80 | 13.00 | 13.60 | 24.90 | 39.40 | 8.70 | 1.80 | | |
| | **AVG** | 1.14 | 1.50 | 0.59 | 0.87 | 1.44 | 1.51 | 2.77 | 4.38 | 0.97 | 0.20 | | |

# Appendix iii

## BONDING BRACKET WITH COMPOSITE CEMENT

| | TOOTH | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 | 9 | 10 | AVG | STDDEV |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | TIME (sec) | TEMPERATURE INCREASE C | | | | | | | | | | | |
| ADT PAC | 10 | 0.8 | 0.6 | 0.6 | 0.8 | 0.7 | 0.5 | 1.5 | 0.9 | 0.9 | 0 | 0.73 | 0.38 |
| | 20 | 1.4 | 1.8 | 1 | 1.3 | 1.8 | 1.4 | 5.4 | 2 | 0.8 | 0.4 | 1.73 | 1.38 |
| | 40 | 2.4 | 4 | 2.2 | 4.9 | 5.8 | 4.4 | 8 | 7.1 | 0.9 | 0.4 | 4.01 | 2.54 |
| POWER SLOT | 5 | 0.3 | 1.7 | 0.2 | 0.9 | 0.3 | 2.3 | 1.6 | 0.5 | 0.1 | 0.1 | 0.80 | 0.79 |
| | 10 | 0.9 | 6.3 | 0.4 | 1.6 | 2.1 | 0.8 | 3.7 | 0.7 | 0.2 | 0.1 | 1.68 | 1.95 |
| | 20 | 0.1 | 0.6 | 2.7 | 5.8 | 1.1 | 3.8 | 6.6 | 1.1 | 0.4 | 0.1 | 2.23 | 2.41 |
| OPTILUX | 10 | 1.1 | 0.9 | 1.9 | 1.8 | 0.9 | 0.4 | 1.8 | 0.4 | 0.1 | 0.1 | 0.94 | 0.70 |
| | 20 | 2.6 | 1.3 | 2.3 | 3.1 | 1.4 | 1.4 | 6.6 | 0.8 | 3.5 | 0.1 | 2.31 | 1.84 |
| | 40 | 5.2 | 3.4 | 7 | 4.3 | 3.3 | 8.3 | 8.4 | 1.8 | 2 | 0.1 | 4.38 | 2.83 |
| TOTAL | | 14.80 | 20.60 | 18.30 | 24.50 | 17.40 | 23.30 | 43.60 | 15.30 | 8.90 | 1.40 | | |
| AVG | | 1.64 | 2.29 | 2.03 | 2.72 | 1.93 | 2.59 | 4.84 | 1.70 | 0.99 | 0.16 | | |

# Appendix vi

**Fisher's PLSD for Baseline ADT**
**Effect: Category for Baseline ADT**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -1.040 | .301 | <.0001 | S |
| 10, 40 | -2.000 | .301 | <.0001 | S |
| 20, 40 | -.960 | .301 | <.0001 | S |

**Scheffe for Baseline ADT**
**Effect: Category for Baseline ADT**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -1.040 | .390 | .0002 | S |
| 10, 40 | -2.000 | .390 | <.0001 | S |
| 20, 40 | -.960 | .390 | .0003 | S |

**Bonferroni/Dunn for Baseline ADT**
**Effect: Category for Baseline ADT**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -1.040 | .394 | <.0001 | S |
| 10, 40 | -2.000 | .394 | <.0001 | S |
| 20, 40 | -.960 | .394 | <.0001 | S |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

**Fisher's PLSD for Baseline XTIP**
**Effect: Category for Baseline XTIP**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -.680 | .129 | <.0001 | S |
| 10, 40 | -1.120 | .129 | <.0001 | S |
| 20, 40 | -.440 | .129 | <.0001 | S |

**Scheffe for Baseline XTIP**
**Effect: Category for Baseline XTIP**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -.680 | .167 | <.0001 | S |
| 10, 40 | -1.120 | .167 | <.0001 | S |
| 20, 40 | -.440 | .167 | .0002 | S |

**Bonferroni/Dunn for Baseline XTIP**
**Effect: Category for Baseline XTIP**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -.680 | .169 | <.0001 | S |
| 10, 40 | -1.120 | .169 | <.0001 | S |
| 20, 40 | -.440 | .169 | <.0001 | S |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

**Fisher's PLSD for Baseline OPTILUX**
**Effect: Category for Baseline OPTILUX**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -.700 | .860 | .0972 | |
| 10, 40 | -1.820 | .860 | .0012 | S |
| 20, 40 | -1.120 | .860 | .0169 | S |

**Scheffe for Baseline OPTILUX**
**Effect: Category for Baseline OPTILUX**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -.700 | 1.113 | .2320 | |
| 10, 40 | -1.820 | 1.113 | .0040 | S |
| 20, 40 | -1.120 | 1.113 | .0487 | S |

**Bonferroni/Dunn for Baseline OPTILUX**
**Effect: Category for Baseline OPTILUX**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| 10, 20 | -.700 | 1.124 | .0972 | |
| 10, 40 | -1.820 | 1.124 | .0012 | S |
| 20, 40 | -1.120 | 1.124 | .0169 | |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

# Appendix v

**Fisher's PLSD for Tooth Only: Time #1**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| ADT, XTIP | .300 | .281 | .0378 | S |
| ADT, OPTILUX | .080 | .281 | .5573 | |
| XTIP, OPTILUX | -.220 | .281 | .1174 | |

**Scheffe for Tooth Only: Time #1**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .300 | .357 | .1089 |
| ADT, OPTILUX | .080 | .357 | .8377 |
| XTIP, OPTILUX | -.220 | .357 | .2837 |

**Bonferroni/Dunn for Tooth Only: Time #1**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .300 | .353 | .0378 |
| ADT, OPTILUX | .080 | .353 | .5573 |
| XTIP, OPTILUX | -.220 | .353 | .1174 |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

**Fisher's PLSD for Tooth Only: Time #2**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .300 | .439 | .1685 |
| ADT, OPTILUX | .170 | .439 | .4268 |
| XTIP, OPTILUX | -.130 | .439 | .5419 |

**Scheffe for Tooth Only: Time #2**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .300 | .557 | .3773 |
| ADT, OPTILUX | .170 | .557 | .7228 |
| XTIP, OPTILUX | -.130 | .557 | .8259 |

**Bonferroni/Dunn for Tooth Only: Time #2**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .300 | .552 | .1685 |
| ADT, OPTILUX | .170 | .552 | .4268 |
| XTIP, OPTILUX | -.130 | .552 | .5419 |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

**Fisher's PLSD for Tooth Only: Time #3**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| ADT, XTIP | 1.690 | 1.076 | .0040 | S |
| ADT, OPTILUX | .160 | 1.076 | .7584 | |
| XTIP, OPTILUX | -1.530 | 1.076 | .0079 | S |

**Scheffe for Tooth Only: Time #3**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| ADT, XTIP | 1.690 | 1.366 | .0142 | S |
| ADT, OPTILUX | .160 | 1.366 | .9525 | |
| XTIP, OPTILUX | -1.530 | 1.366 | .0267 | S |

**Bonferroni/Dunn for Tooth Only: Time #3**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| ADT, XTIP | 1.690 | 1.352 | .0040 | S |
| ADT, OPTILUX | .160 | 1.352 | .7584 | |
| XTIP, OPTILUX | -1.530 | 1.352 | .0079 | S |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

# Appendix iv

**Fisher's PLSD for Bracket: Time #1**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | -.070 | .544 | .7898 |
| ADT, OPTILUX | -.210 | .544 | .4276 |
| XTIP, OPTILUX | -.140 | .544 | .5951 |

**Scheffe for Bracket: Time #1**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | -.070 | .690 | .9641 |
| ADT, OPTILUX | -.210 | .690 | .7236 |
| XTIP, OPTILUX | -.140 | .690 | .8648 |

**Bonferroni/Dunn for Bracket: Time #1**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | -.070 | .683 | .7898 |
| ADT, OPTILUX | -.210 | .683 | .4276 |
| XTIP, OPTILUX | -.140 | .683 | .5951 |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

**Fisher's PLSD for Bracket: Time #2**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .050 | 1.213 | .9320 |
| ADT, OPTILUX | -.580 | 1.213 | .3285 |
| XTIP, OPTILUX | -.630 | 1.213 | .2896 |

**Scheffe for Bracket: Time #2**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .050 | 1.540 | .9963 |
| ADT, OPTILUX | -.580 | 1.540 | .6121 |
| XTIP, OPTILUX | -.630 | 1.540 | .5619 |

**Bonferroni/Dunn for Bracket: Time #2**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | .050 | 1.524 | .9320 |
| ADT, OPTILUX | -.580 | 1.524 | .3285 |
| XTIP, OPTILUX | -.630 | 1.524 | .2896 |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

**Fisher's PLSD for Bracket: Time #3**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value | |
|---|---|---|---|---|
| ADT, XTIP | 1.670 | 1.635 | .0458 | S |
| ADT, OPTILUX | -.370 | 1.635 | .6403 | |
| XTIP, OPTILUX | -2.040 | 1.635 | .0173 | S |

**Scheffe for Bracket: Time #3**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | 1.670 | 2.076 | .1288 |
| ADT, OPTILUX | -.370 | 2.076 | .8938 |
| XTIP, OPTILUX | -2.040 | 2.076 | .0545 |

**Bonferroni/Dunn for Bracket: Time #3**
**Effect: Category for Tooth Only: Time #1**
**Significance Level: 5 %**

| | Mean Diff. | Crit. Diff | P-Value |
|---|---|---|---|
| ADT, XTIP | 1.670 | 2.054 | .0458 |
| ADT, OPTILUX | -.370 | 2.054 | .6403 |
| XTIP, OPTILUX | -2.040 | 2.054 | .0173 |

Comparisons in this table are not significant unless the corresponding p-value is less than .0167.

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