Comparing the Degree of Exothermic Polymerization in Commonly Used Acrylic and Provisional Composite Resins for Intraoral Appliances

Rice CA, DVM; Jessica Riehl, DVM; Karl Broman, PhD; Jason W. Soukup, DVM

Summary:

The use of dental acrylics and composite resins in veterinary dentistry has become widespread. However, their use is not without potential complications. All acrylics and composite resins produce an exothermic reaction during the polymerization process. The aim of the current study was to evaluate thermal conduction during the polymerization reaction of each material to offer clinical guidelines when choosing a material with particular consideration for the significant volumes typically used. Results showed that methylmethacrylate based resins generated a significantly higher degree of heat during polymerization. Bis-acryl based composite resins generated a significantly lower degree of heat during polymerization, making them the material of choice to potentially minimize thermal injury to the dentin-pulp complex. It is the responsibility of the clinician to become aware of all materials available, and to have an understanding of their properties to guide them in making sound clinical judgments. ***J Vet Dent 29 (2); #$% - $&*, 2012***

Introduction

Methacrylate based resins have had multiple applications for their use in the medical field since their discovery almost 70 years ago.[1] Two of the most common uses for acrylic and composite resins in veterinary dentistry are the intraoral application of inclined planes to correct linguoversion of mandibuar canine teeth and the application of intraoral splints to aid in the repair of maxillary and mandibular jaw fractures (Fig. 1).[2,3] These materials have gained widespread use. However, their use is not without potential complications. It is well known that the polymerization reaction produces an exothermic release of energy. There is concern when using resins in the oral cavity since direct contact with the teeth has the potential to cause pulpitis and eventual pulpal necrosis. Several studies have attempted to assess the degree of thermal conduction of resins to the teeth.[4-6] However, these studies have been based on the extraoral fabrication of prosthetic crowns, bridges, and fixed partial dentures in humans. The concern for exothermic injury to teeth in veterinary dentistry may also be related to the quantity (or volume) of material used and the possibility of a greater exothermic reaction and subsequent thermal damage to dental tissues. The heat released by these materials may also cause injury to adjacent soft tissues of the oral cavity. In addition, in veterinary patients, it is likely that all acrylic and composite resin-based appliances created intraorally will predispose the patient to some degree of gingivitis or mucositis.

No ideal acrylic or composite material exists in dentistry. However, comparison of characteristics between materials can lead to a better understanding of the compromises one must accept when choosing a material; such as ease of handling and esthetics compared with strength and heat released during polymerization. The purpose of the present study was to compare the exothermic potential of acrylic and composite resins commonly used for intraoral application in veterinary dentistry.

Materials and Methods

Five samples each of four different materials were used for comparison against one another to determine the maximum temperature rise and the temporal temperature profiles of each material at 5° C, 11° C, and16° C above body temperature during polymerization of each material in a controlled, repeatable laboratory setting. The four materials consisted of a chemically cured acrylic resin[a] (Ortho-Jet), a chemically cured provisional composite resin[b] (Maxi-Temp), a dual-cure provisional composite resin[c] (TempSpan), and a chemically cured fiber-reinforced provisional composite resin[d] (Build-It). Each material was prepared using the manufacturers specified suggestions for application. A 10-ml plastic graduated cylinder was used to create a 13.6-cm X 11-mm long cylindrical mold. A calibrated k-type thermocouple with an accuracy of 0.1° C was used to measure the internal temperature of the resin during polymerization. The k–type thermocouple probe was placed in the center of the open end of the graduated cylinder to a depth of 2.5-cm and temperature readings recorded at 15-second intervals (Fig. 2). Temperature recordings began within 15-seconds of first immersing the probe into the poured material and recorded through peak temperature and until the internal temperature of the mold had reduced to body temperature (38° C).

The acrylic resin was prepared using the manufacturers suggested "salt and pepper" technique until the graduated cylinder was completely filled using a mixing ratio of 5:1 powder to liquid ensuring a homogenous mix of the polymethylmethacrylate beads of the polymer powder and the liquid of the methylmethacrylate monomer. Each of the provisional composite resins was delivered into the graduated cylinder using a compatible composite delivery syringe and appropriate mixing tip. Modifications were made to the mixing tips to ensure that the material would reach the bottom of the graduated cylinder without slumping along the sides and creating air voids during filling. In order to do this, a 1 ml tuberculine syringe without needle and plunger was fitted to the end of the mixing tip. This made the tip long enough to reach the bottom of the graduated cylinder. During delivery, the tip was withdrawn from the bottom of the graduated cylinder as the composite resin was expressed from the delivery syringe and the graduated cylinder began to fill. This was done slowly and carefully to ensure an adequate void-free fill throughout. Once filled, the k-type tem-

perature probe was inserted into the open end of the graduated cylinder as described previously. Temperature readings were taken in 15-second intervals. Five molds were made for each material with temperature readings taken from each material tested.

Four separate parameters for each material were analyzed. The outcomes of maximum temperature, as well as the time each material spent at 5° C, 11° C, and 16° C above body temperature were evaluated. For each outcome a one-way analysis of variance (ANOVA) was performed between materials to test for differences in the mean maximum temperature as well as in the average time each material spent at 5° C, 11° C, and 16° C above body temperature. Following a significant F-test, means were compared pair-wise using the Tukey Honest Significant Difference adjustment. Statistical significance was set at $P < 0.05$.

Results

The maximum temperature as well as the time to complete the exothermic effect of the polymerization reaction for all samples of each material tested were recorded (Fig. 3). There were significant differences among the four materials tested for all four outcomes (maximum temperature reached and amount of time each material spent at or over 5° C, 11° C, 16° C above body temperature) with P-values < 0.001 for all cases. The Maxi-Temp showed the lowest values for all four outcomes, thus making it the material of choice when risk of thermal injury to the pulp is of concern. Ortho-Jet showed a significantly higher maximum temperature than the other three materials, but for time spent at 5° C, 11° C, or 16° C above body temperature, the Ortho-Jet, TempSpan, and Build-It materials were not significantly different. TempSpan reached maximum temperature most quickly, Ortho-Jet acrylic demonstrated the highest maximum temperature, and Maxi-Temp demonstrated the lowest maximum temperature (Fig. 4). The methylmethacrylate based resin (Ortho-Jet) produced higher maximum temperature rises than either the bis-acryl based (Maxi-Temp) or bis-GMA based (TempSpan and Build-It) materials. The bis-acryl based composite resin produced a significantly lower maximum temperature rise than the bis-GMA based composite resins. There was no significant difference in the maximum temperature rise of the two bis-GMA based composite resins.

Discussion

Provisional dental materials fall into two basic categories based on their chemical composition: methacrylates, more commonly known as acrylics or acrylic resins, and provisional composite resins.[7] Methacrylates can be divided further into methylmethacrylates, ethylmethacrylates, and vinylmethacrylates.[7,8] Acrylic resins, like the Ortho-Jet acrylic used in this study, are typically composed of a powder-liquid formulation; the powder, composed of small grains of polymer, and the liquid monomer, consisting of a methacrylate alone or in conjunction with other methacrylate-type monomers. An amine activator or accelerator is usually added to the monomeric component of the material to aid the polymerization reaction along with an inhibitor, which is added to prevent polymerization during storage and to prolong the shelf life of

Figure 1

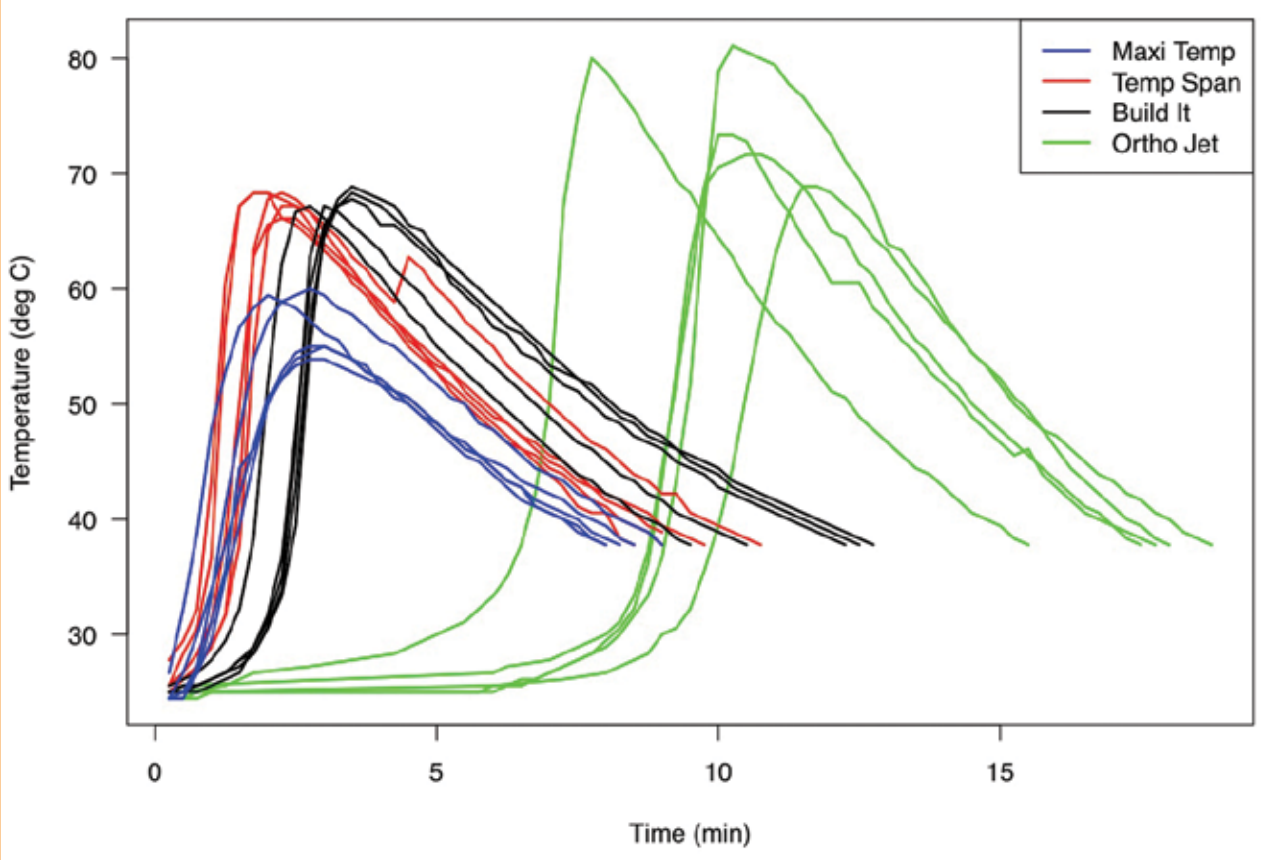
Photograph showing the typical appearance of an intraoral splint made from a provisional composite resin fabricated intraorally in a dog.

Figure 2

Photograph showing a K-type thermocouple and probe inserted 2.5-cm into the open end of a 10-ml plastic graduated cylinder filled with a provisional composite resin.

Figure 3

Graph showing temperature rise and time of exothermic effect of the polymerization reaction of all samples of four different materials tested. Note the sudden spike in temperature of one sample of TempSpan at 4:15 to 4:45. This was considered artifactual.



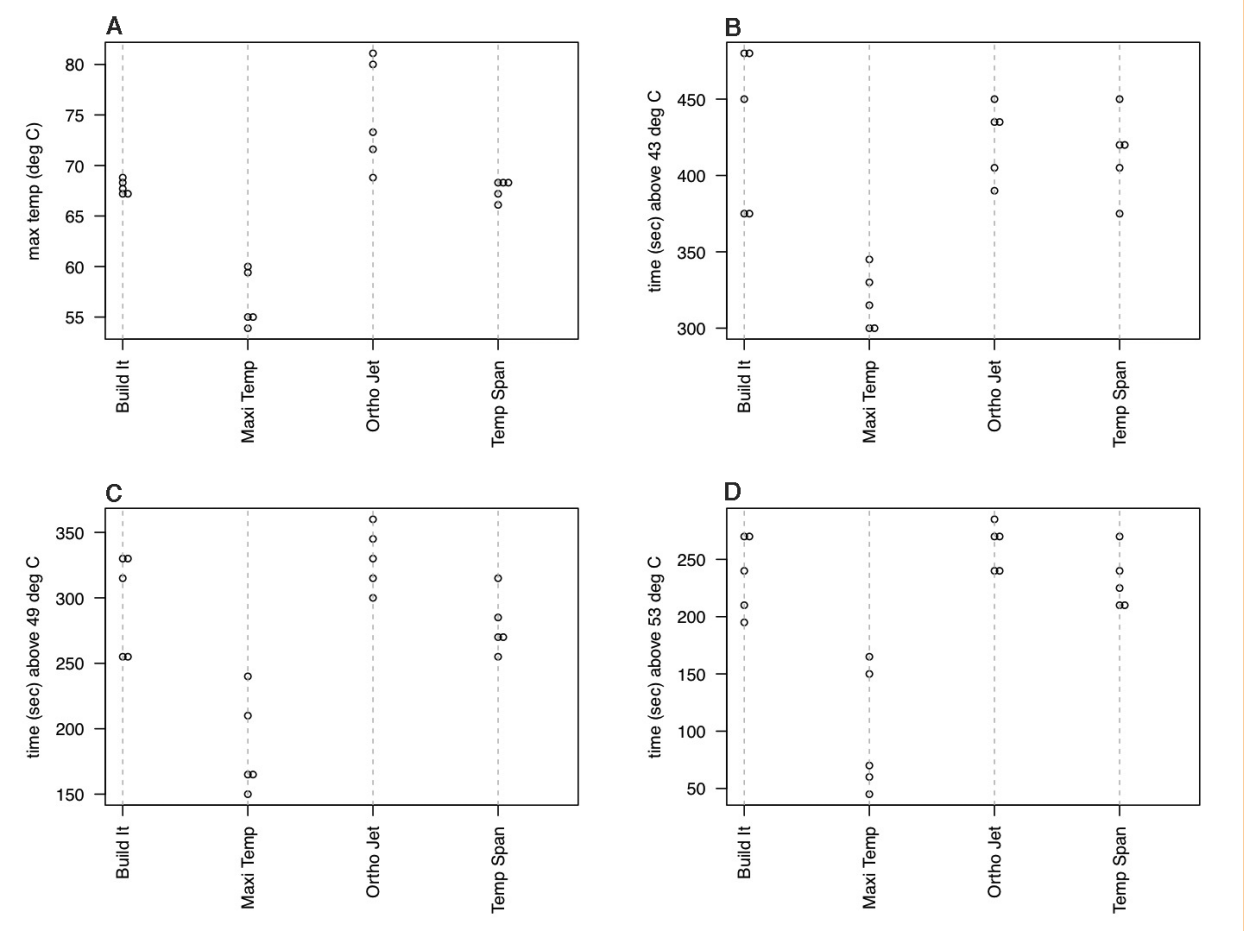
the material.[9] In the current study, methylmethacrylate based resins produced significantly higher maximum temperatures than any other bis-acryl or bis-GMA based resin tested. Aside from the disadvantage of being the most exothermic material tested, another limitation to the use of these materials includes the need for preparation in a well ventilated room or under a ventilation hood in a laboratory setting due to the significant toxic fumes released during mixing.

Provisional composite resins like Maxi-Temp, TempSpan and Build-It are similar in composition to dental composite restoratives. The monomeric components of dental composites are called dimethacrylates.[7,9] The most common dimethacrylates used in provisional composites are bis-acryl, bis-GMA, or urethane methacrylate.[7,9] The self, or chemical curing systems are typically packaged in two separate cylinders within a plastic cartridge that is compatible with its own delivery syringe and mixing tip. An initiator, typically benzoyl peroxide, along with the composite resin is added to one cylinder in the cartridge and an activator mixed with composite resin is added to the other.[8] When mixed, the tertiary amine in the activator causes the initiator to become reactive, or "initiate" reactivity of the monomeric component, which reacts with other monomers in the material to begin polymerization.[8] Some advantages encountered during the preparation of each mold in this study included ease of handling, no release of toxic volatile gases, and consistent homogeneity of the resultant product when it is expressed from the mixing tip. This is evident where the groupings among individual samples tested were much tighter for the composite resins than the methacrylate based resin (Fig. 3).

The first step of the polymerization reaction commonly observed in dental materials involves activation of the initiator. Activation can be accomplished by three different energy sources: heat, chemical, and light.[9,10] Chemical activation is common in most modern provisional self-cure acrylics and composites, although some may be light or duel-cured.[9,10] Heat activated materials are typically limited to use in the laboratory setting. After the initiator reacts with the carbon-carbon double bond of the monomer, a carbon-carbon single bond is formed along with an unpaired electron, known as a free radical. The resultant free radical is highly reactive and able to react with another carbon-carbon double bond of the monomer. This causes yet another split of the carbon-carbon double bond, transferring the free radical to it from the previous monomer, which is then allowed to react with another monomer. The splitting of the carbon-carbon double bond during initiation results in the release of heat. The resultant heat then accelerates the

Figure 4

Plots of all samples of each material tested including maximum temperature (A) and time each sample spent at or over 5° C (B), 11° C (C), and 16° C (D) above body temperature.



polymerization of the monomer thus giving off more heat.[9,10]

Previous studies have demonstrated that 15 % of healthy pulps failed to recover from intrapulpal temperature rises of 5.55° C above body temperature.[4,11] Further, 60 % of pulps failed to recover from a temperature rise of 11.1° C above body temperature and 100 % of pulps were incapable of recovering from intrapulpal temperature rises of 16.65° C above body temperature, leading to eventual irreversible pulpitis and pulpal necrosis. In this latter study, a soldering iron in contact with the enamel was used to induce intrapulpal temperature rises and a very small thermistor encased in a 22-gauge needle, introduced into the pulp chamber through a small hole in the crown, was used to measure temperature changes. Other studies reported similar findings.[12,13] Interestingly, no differences were observed between Ortho-Jet acrylic, TempSpan and Build-It with regards to the amount of time each material spent at either 5° C, 11° C, or 16° C above body temperature despite differences in the monomers of these materials and method of preparation. Maxi-Temp had the lowest values for all four parameters tested.

Non-diseased human teeth have been shown to be able to withstand temperatures between -7° C and 75° C without damage to pulpal tissue.[14] In this study of non-diseased human teeth, a differential calorimeter was inserted into the pulp, which was capable of measuring temperature changes during polymerization in real time. It has also been reported that pulps exposed to temperatures greater than 5° C above body temperature for 1 minute caused odontoblastic death.[15] In addition, previous studies have reported heat-induced osteonecrosis in a rabbit model after exposing bone to temperatures of 50-55° C for 30-seconds to 3-minutes.[16,17] Reports like these indicate that not only is maximum temperature reached a factor in predisposing to cell injury and death but the duration at which these temperatures persist also seem contributory. For this reason, the temporal temperature profiles of all four materials within the present study were tested at similar parameters (Fig. 4).

It had also been reported that an intrapulpal rise in temperature of 2.3° C caused minimal pulpal changes but at 5.6° C above body temperature, a remarkable pulpal response was observed.[11] Within approximately 2-months, 85 % of teeth can recover from the inflammatory reaction caused by the 5.6° C

above body temperature increase.[4] It has been concluded in people that the maximum tolerable temperature for drinking liquids is between 50-55° C.[18] Teeth exposed to temperatures of liquids in that range produced a rise in temperature at the tooth's surface to 47° C.[18] Also, *in-vitro* tests showed tooth surfaces exposed to temperatures of 50° C for 5-seconds produced intrapulpal temperature rises of 0.9° C above body temperature, well within the dental pulp's ability to recover from such insults.[18] In this latter study, intrapulpal temperature rise was measured *in-vitro* using an extracted human premolar tooth. The apex of a root of the tooth was resected and a thermistor bead was placed retrograde into the pulp chamber.

The thermal conductivity of human dentin has been determined in previous studies.[19,20] With this information, it is theoretically possible to determine intrapulpal temperature increases based on subjecting a tooth's surface to a given temperature. One limitation of the current study is that it is not an *in vivo* study and an objective correlation between an exothermic release of energy during the polymerization reaction of a given material and an intrapulpal temperature increase could not be determined, however some correlation needs to be drawn. In addition, results can vary significantly depending on the age of the patient as dentinal thickness increases significantly within the first 4-years of life in the dog.

Dispersal of heat from an exothermic reaction has been taken into account in some studies simulating intraoral wet conditions and pulpal blood flow.[6] Teeth have a unique response that leaves them vulnerable to thermal energy and unable to respond favorably.[6] An increase in intrapulpal temperature does not increase pulpal blood flow. Rather, as a consequence of inflammation, swelling results in decreased blood flow.[6] A method to potentially decrease the amount of heat conduction to dental tissues during polymerization would be to lavage the material with cool water from a three-way air-water syringe until heat release is no longer a factor. This may make the decision as to which material to choose less restrictive.

Most studies performed previously evaluating the exothermic reactions of dental materials have used significantly smaller quantities of material than we considered here. By using different sized samples, it was determined that the maximum temperature reached was directly proportional to the volume of material.[21] Sample volumes for these studies were 750 mm^3, 1500 mm^3, and 3000 mm.[3-21] These samples were of a flat, rectangular shape and the study acknowledged that surface area over which heat can dissipate is greater for a flat sheet of acrylic versus a more spherical shape. In veterinary dentistry, given the shape of the canine and feline dental anatomy, rarely is a flat shape of material formed. In the current study we attempted to use shapes (cylinder) and volumes consistent with that typically used in veterinary dental practice. Our mean sample volume was 51,718 mm^3; 17-70 times greater than the volumes used previously.[21]

All current literature supports the concept that an exothermic reaction occurs during the polymerization reaction of all acrylic and composite resins. Unfortunately, studies cannot be easily compared against one another due to differing methods and materials used. In one study, mean intrapulpal temperatures increased 37.76 - 39.4° C when simulating a provisional crown fabrication.[4] Another study reported temperatures increasing between 33.3° C and 53.3° C.[22] However, the volumes of material used in this latter study were much greater than those described in the former and tests were not conducted in a wet environment of 36° C. In comparison, in the current study a sample volume of 51,718 mm^3 was used to evaluate exothermic energy release during polymerization of four different acrylic and composite resins. The results of this study showed temperature increases between 55 - 81.1° C. These results are significantly higher than those reported in previous studies and consistent with the observation that maximum temperatures increased with greater material volume.

One study showed a mean intrapulpal increase in temperature ranging from 3.95 - 6.06° C regardless of the material tested.[6] Of those materials, methacrylate based acrylic resins had the greatest intrapulpal increases in temperature and the bis-acryl composite-based resins had the lowest temperature increase.[6] These results are consistent with the results of the study reported here.

The chemical composition of a material affects the temperature reached at peak polymerization, thus validating the need for comparison between materials.[21] In addition, curing method has been related to thermal reaction.[22] Chemically cured methylmethacrylates exhibit a temperature increase significantly higher than light-cured or chemically cured bis-acryl composite resins.[4-6] The results of our study reflect those of previously cited studies. In the study reported here, direct correlations between maximum temperature rise of the material, the time these materials spent at 5° C, 11° C, 16° C above body temperature, and intrapulpal temperature rise could not be made. However, it seems logical that materials that exert the highest degree of temperature rise for the longest time are more likely to cause pulpal injury. This further proves that a study of the applications of these materials for use in veterinary dentistry directly correlating exothermic energy release and intrapulpal temperature increase is necessary.

The use of acrylic and provisional composite resins in veterinary dentistry has gained widespread use. However, their use is not without potential complications. During polymerization, all acrylic and provisional composite resins have the potential to cause irreversible pulpitis and pulpal necrosis. This study showed that methylmethacrylate based acrylic resins produce a significant higher and longer exothermic reaction than bis-acryl or bis-GMA based provisional composite resins in volumes consistent with their use in veterinary dentistry. Bis-acryl based provisional composite resins produced significantly lower and shorter exothermic reactions than any other material tested, making them the material of choice when thermal injury to the pulp is of concern. There are many materials on the market available to veterinarians. It is the responsibility of the clinician to become aware of all materials available to them, as well as have an understanding of their properties to guide them in making sound clinical judgments.

---

[a] Ortho-Jet (Polymethylmethacrylate), Lang Dental Manufacturing, Wheeling, IL. Batch 1334-10AT

[b] Maxi-Temp (bis-acryl), Henry Schein Inc., Melville, NY. Batch 633516

[c] TempSpan (bis-GMA), Pentron Clinical Technologies, Wallingford, CT. Batch 203904

[d] Build-It (bis-GMA, fiber reinforced), Pentron Clinical Technologies, Wallingford, CT. Batch 204788

Author Information

From the University of Wisconsin-Madison, School of Veterinary Medicine, 2015 Linden Drive; and the Department Biostatistics and Medical Informatics (Broman), 1300 University Avenue, Rm 4710 MSC, Madison, WI, 53706. Dr. Rice's current address is the University of Pennsylvania, School of Veterinary Medicine, 3900 Delancey Street Philadelphia, PA, 19102. Email: carice@vet.upenn.edu

References

1. Laing, PG. Clinical experience with prosthetic materials: historical perspectives, current problems, and future directions. In: Syrett BC, Acharya A. *Corrosion and degradation of implant materials.* American society for testing and materials, 1979; 199-211.

2. Niemiec B. Intraoral splint application. *J Vet Dent* 2003; 2:123-126.

3. Legendre LF. Building a telescopic inclined plane intraorally. *J Vet Dent* 2010; 1:62-65.

4. Michalakis K, Pissiotis A. Comparison of temperature increase in the pulp chamber during the polymerization of materials used for the direct fabrication of provisional restorations. *J Prosth Dent* 2006; 6:418-423.

5. Moulding MB, Teplitsky PE. Intrapulpal temperature during direct fabrication of provisional restorations. *Intl Jour Prosth* 1990; 3:299-304.

6. Whalen S, Bouschlicher M. Intrapulpal temperature increases with temporary crown and bridge material. *Gen Dent* 2003; 6:534-537.

7. Ferracane JL. Provisional restoratives. In: Ferracane JL. *Materials in dentistry, principles and applications*, 2nd ed. Lippencott, Williams and Wilkins, 2001; 223-235.

8. Rawls RH. Dental polymers. In: Anusavice. *Phillips' science of dental materials*, 11th ed. St. Louis: Saunders, 2003; 143-169.

9. Ferracane JL. Direct aesthetic anterior restoratives. In: Ferracane JL. *Materials in dentistry, principles and applications*, 2nd ed. Lippencott: Williams and Wilkins, 2001; 85-118.

10. Ferracane JL. Polymeric materials: the basics. In: Ferracane JL. *Materials in dentistry, principles and applications*, 2nd ed. Lippencott: Williams and Wilkins, 2001; 255-280.

11. Zach L, Cohen G. Pulp response to externally applied heat. *Oral surg oral med oral pathol* 1965; 19:515-310.

12. Lieu C, Nguyen TM, Pavant L. In vitro comparison of peak polymerization temperature of 5 provisional restoration resins. *J Can Dent Assoc* 2001; 1:36-39.

13. Stanley H. Pulpal response to dental techniques and materials. *Dental clinics of north america* 1971; 1:115-126.

14. Chirtoc M, Bicanic D. Monitoring the polymerization process of acrylic resins. *Intl Jour Prosth* 1995; 3:259-264.

15. Peters O, Peters Cleaning and shaping of the root canal system. In: Cohen S, Burns RC editors. *Pathways of the pulp*, 9th ed. St. Louis: Mosby, 2006; 290-357.

16. Eriksson AR, Albrektsson T. Temperature threshold levels for heat-induced bone tissue injury: A vital-microscopic study in the rabbit. *J Prosth Den* 1983; 1:101-107.

17. Lundskog J. *An experimental investigation of the thermal properties of bone tissue and threshold levels for thermal injury.* Thesis 1972; University of Goteborg.

18. Plant CG, Jones DW. The heat evolved and temperatures attained during setting of restorative materials. *Brit Dent J* 1974; 6:233-238.

19. Braden M. Heat Conduction in normal human teeth. *Archs Oral Biol* 1964; 4:479-486.

20. Fanibunda KB, De Sa A. Thermal conductivity of normal and abnormal human dentine. *Archs Oral Biol* 1975; 7:457-459.

21. Vallittu PK. Peak temperatures on some prosthetic acrylates on polymerization. *J Oral Rehab* 1996; 11:776-781.

22. Driscoll CF, Woolsey G, Ferguson WF. Comparison of exothermic release during polymerization of four materials used to fabricate interim restorations. *J Prosth Dent* 1991; 4:504-506.