Acrylic Resin Polymerization in Direct Contact to the Abutment and the Temperature at Bone-Implant Interface: A Pilot In Vitro Study

Mahmood Kazemi, DMD, MS[1]
Hamid Jalali, DMD, MS[1]
Mehrdad Eghtedari, DMD, MS[2]
Roozbeh Sadrimanesh, DDS, MS[3]*
Pooyan Sadr-Eshkevari, DDS[4]
Peter Maurer, Dr Med, Dr Dent, PD[5]

Three autopolymerizing acrylic resins were applied to a titanium alloy abutment connected to 2 different diameters of an implant. The implants were embedded in fresh iliac bone of sheep in a 37°C water bath. Temperature changes were recorded via embedded thermocouples at the cervical (T1) and apical (T2) regions of the implant surface. Polymerization temperature of acrylic resins did not seem to exceed the critical threshold of 47°C.

Key Words: bone regeneration, dental implant, implant-supported dental prosthesis, osseointegration, polymerization temperature, polymethacrylic acids

INTRODUCTION

Autopolymerizing acrylic resins, which are routinely used in the fabrication of implant-borne temporary prostheses, are associated with high polymerization temperatures that might pose a potential risk to the bone-implant interface when directly applied to the implant abutments.[1] The mean polymerization temperatures of 4 common autopolymerizing acrylic resins is reported to be between 33°C and 55°C.[2,3] Excess temperature may impair the regenerative potential of the bony tissue and result in necrosis (thermal damage).[4,5] Studies have shown that bony changes in response to temperature start at 47°C. Temperatures equal to or above 60°C can cause permanent vascular stasis and bone necrosis.[6,7]

Several studies have concerned the temperature rise and its effects during bone drilling.[8–11] Several others have concerned the exothermal activity of polymerizing material and the effect of different factors on temperature rise.[12–16] It has been shown that successful implant treatment depends on the proper integration between the implant and the surrounding bony structures and soft tissues.[17]

To the best of our knowledge, however, few studies have addressed the effect of temperature rise during acrylic resin polymerization in contact with the implant abutment.[1] This study was then conducted to assess the temperature alterations of implant surface during the polymerization of 3 acrylic resins in direct contact with the abutments placed in bone. The null hypothesis stated that the

[1] Department of Prosthodontics, School of Dental Medicine, Tehran University of Medical Sciences, Tehran, Iran.
[2] Department of Prosthodontics, School of Dental Medicine, Ahvaz Jondishapoor University of Medical Sciences, Ahvaz, Iran.
[3] Department of Oral and Maxillofacial Prosthodontics and Implants, School of Dental Medicine, Tehran University of Medical Sciences, Tehran, Iran.
[4] Farzan Clinical Research Institute, Tehran, Iran; Department of Oral and Maxillofacial Plastic Surgery, Ruhr Universität, Bochum, Germany.
[5] Department of Oral and Maxillofacial Surgery, Ruhr Universität
* *Corresponding author*, e-mail: sadrimnsh@yahoo.com
DOI: 10.1563/AAID-JOI-D-10-00067

maximum temperature at the bone-implant interface does not reach or exceed the critical threshold of 47°C.

Materials and Methods

Implants and thermometer

In the present study, 20 SLA-coated ITI (solid-screw) implants (ITI, Straumann, Basel, Switzerland), 10 mm in length and 4.8 mm in platform diameter, were used. Ten implants were 3.3 mm in diameter and the other 10 were 4.1 mm in diameter. A thermometer (TES-1307 Datalogging K/J, TES Inc, Taiwan) with the accuracy of 0.1°C was utilized along with the standard J-type, ferro-constantan thermocouples to record the temperature alterations. To create a flat surface and hence optimize the contact between the flat end of the thermocouple and the fixture body, 2 pits were prepared by cylindrical end-cutting carbide bur on the outer surface of each fixture. The coronal pit (T1) was placed 2 mm under the inferior border of the collar, and the apical one (T2) was placed 2 mm above fixture apex. The pits were 0.5 mm in depth and 1 mm in diameter (equal to the diameter of the thermocouple probe) and were placed on a line parallel to the longitudinal axis of the fixture.

Guide jig

A metal jig was prepared to serve as a guide for proper accessibility to the pits: castable plastic abutment was used to create a customized implant-supported guide jig. A 30-mm-long wax arm was formed perpendicular to the long axis of the implant and attached to the plastic head of the abutment. Another wax arm was prepared equal to the fixture-abutment assembly in length and attached to the free ending of the former arm parallel to the fixture. This apparatus was prepared so that a hypothetical line passing the pits and the 2 arms would be on a flat surface. The distances of T1 and T2 from the highest coronal point on the abutment were 14.8 mm and 20.8 mm, respectively. To locate these points by the jig, 2 similar pits with similar distances from the horizontal arm of the guide jig were drilled out on the parallel wax arm. The apparatus was then casted (Figure 1). To locate the pits after the implant insertion into the bone, a guiding line was cut on the fixtures and then on the abutments.

Bone and implantation

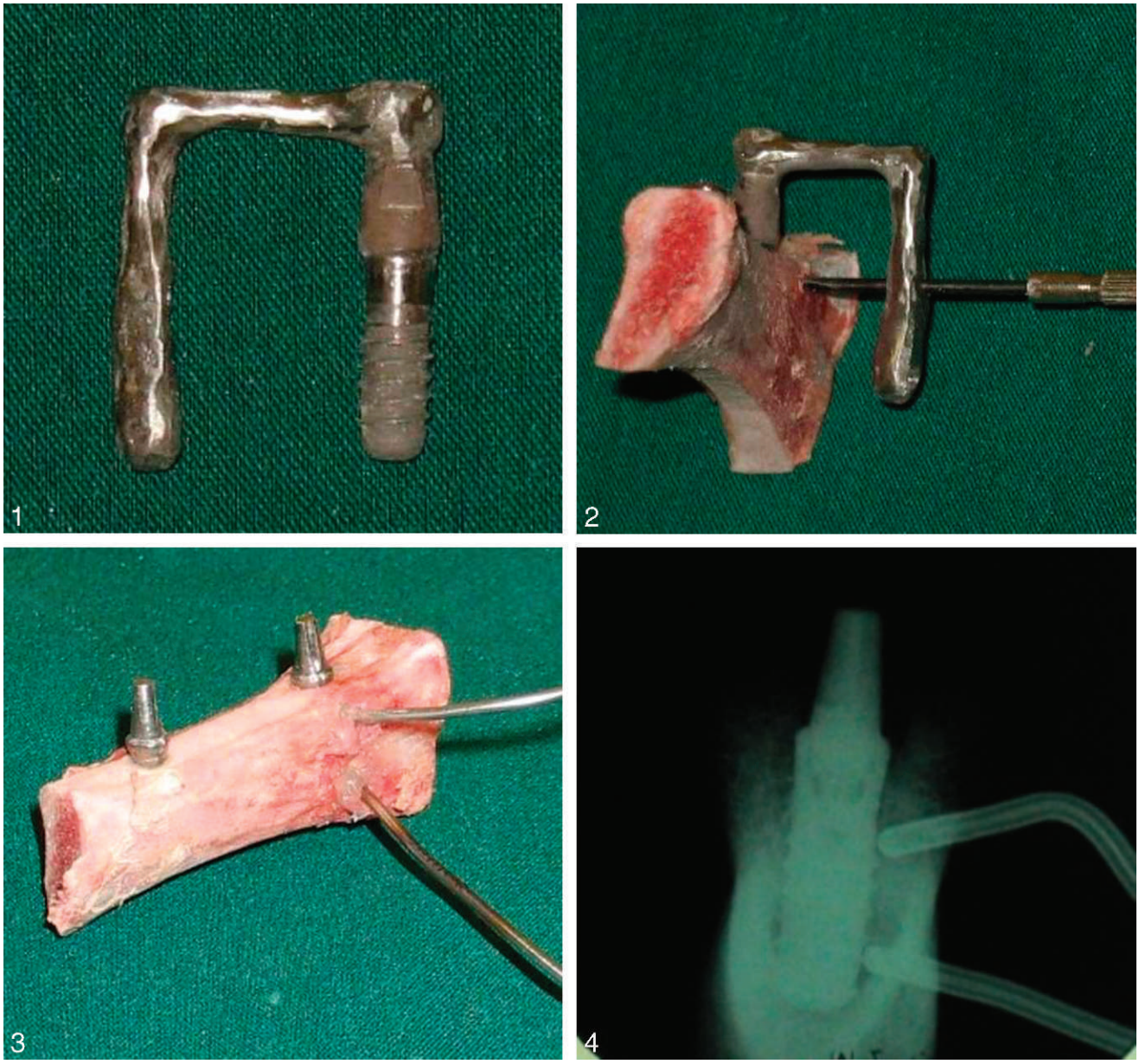
Freshly (within 2 hours before the placement of the implants) sacrificed sheep iliac was used. Ten bony segments (50 × 15 × 20 mm) were prepared. Insertion sites were prepared using ITI system surgical burs based on the manufacturer instructions. At least 1-mm bony margin was saved around the implants in the thinnest areas. Implants were placed so the marked cut of the collar will face the thinnest bony wall. The jig was then fixed on the fixture by abutment screw. Pit-corresponding points were marked on the bone using tissue marker according to the holes on the metal jig (Figure 2). The marked points were drilled to the fixture surface with a 1-mm in diameter carbide bur to place in contact with T1 and T2. The jig was then removed, thermocouple probes were placed, and the correct placement of the tips of the thermocouple probes in the pits was assessed radiographically (Figures 3 and 4). Abutments (5.5-mm height, Solid, ITI) were tightened to the implants.

Clinical simulation

Hollow metal molds (400 mm^3) shaped as mandibular molars were fabricated. Molds were in contact to the fixtures in the shoulder area, and their inner occlusal surface was 2 mm away from the highest coronal points of the abutments. To measure the temperature rise during the polymerization of the acrylic resin, a hole was prepared on the lateral wall of each mold so that the thermocouples could reach the inner room of the molds. The temperature measured on this point was named T3. Each complex was soaked in a water bath so that the water level would not exceed beyond the bone level. The temperature was raised so that the thermometer would show T1 and T2 as equal to 37°C.

Acrylic resins

Three acrylic resins—Duralay (Reliance, packed by Asia ChemiTeb Co), Tempron (GC Corporation, Tokyo, Japan), and Acropars TRII (Marlic Medical Industries Co, Tehran, Iran)—were evaluated in this study. To improve the accuracy of the study, the powder–liquid mix weight percent was considered



Figures 1–4. Figure 1. Guide jig used to locate study points on the bone-embedded implants. **Figure 2.** With the jig in place, tissue marker was used to locate the proper drilling points on the bone to reach T1 and T2. **Figure 3.** Thermocouple probes were placed in contact with T1 and T2. **Figure 4.** Correct contact of the thermocouple probes to the pits was assessed radiographically using an occlusal film

as the mixing index. To weigh the powder, digital balance with the nominal resolution of 0.001 g was used. A pipette with the precision of 0.1 mL was used to measure the liquid volume. The proportion of powder to liquid was determined based on the instructions of the manufacturer. The proportion was 1 g to 0.5 mL for Tempron, 2 g to 1 mL for Duralay, and 1.7 g to 1 mL for Acropars TRII. Powder and liquid were mixed and placed into the mold. The mold was then placed on the abutment and excess resin was removed. A layer of petroleum gel was applied on the probe tip (T3) to prevent its adherence to the acrylic resins. The tip was then inserted into the mold from the side wall.

The thermometer was wired to a computer via RS232 media. Software within the thermometer was able to record the temperature at predetermined time intervals and draw the temperature graph. Since the purpose was not just to determine the time needed to reach the maximum polymerization

TABLE 1

Descriptive statistics of all studied acrylic resins and implant diameters*

Acryl	Implant		Minimum	Maximum	Mean	SD
Acropars	Narrow	T1	.40	.80	.5200	.13166
		T2	.10	.20	.1833	.04082
		T3	40.00	43.60	41.3600	1.04478
	Wide	T1	.40	.90	.5700	.14944
		T2	.10	.30	.2000	.10000
		T3	40.40	45.20	42.0800	1.81402
Duralay	Narrow	T1	.50	1.60	1.0400	.34705
		T2	.20	.50	.2833	.11690
		T3	42.00	45.90	43.3300	1.10760
	Wide	T1	.40	1.30	.8600	.30258
		T2	.10	.20	.1500	.07071
		T3	42.00	48.20	45.1900	2.36899
Tempron	Narrow	T1	.30	1.20	.7600	.32728
		T2	.10	.20	.1500	.07071
		T3	41.00	44.30	42.3500	1.06901
	Wide	T1	.30	1.20	.6900	.30714
		T2	.20	.30	.2500	.07071
		T3	40.90	48.00	44.6800	2.45981

*Three different acrylic resins—Acropars, Duralay, and Tempron—were polymerized on 2 diameters of fixtures, 3.3 mm (narrow) and 4.1 mm (wide). The maximum, minimum, mean, and SD of the temperature rise were recorded as shown.

temperature, temperature alterations were recorded every 2 seconds. This process was carried out until the acrylic resin reached its initial temperature. This procedure was done for the evaluation of 2 implant diameters (3.3 and 4.1 mm) and 3 different acrylic resins. Since the minimum number of specimens to obtain significant results was 10, the temperature alterations were recorded 60 times. The data were classified and statistically analyzed using SPSS v.15 (SPSS Inc, Chicago, Ill). Two-way analysis of variance (ANOVA) and Bonferroni post hoc tests were used to statistically analyze the data.

RESULTS

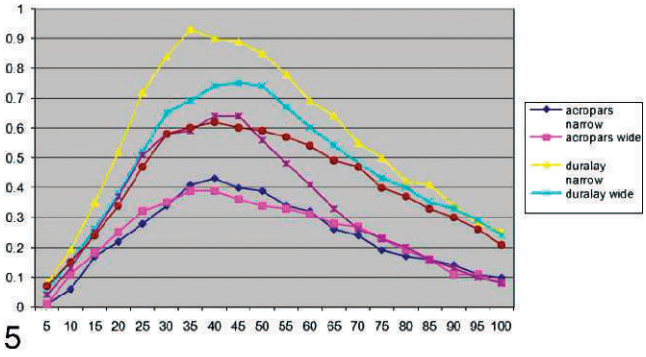
Table 1 summarizes the findings of the present study for each implant diameter and each acrylic resin type. Based on this table, the maximum temperature rise occurred in T1 as 1.6°C with an average of 1.04°C in the Duralay and narrow implant group. The minimum temperature rise occurred in T1 as 0.8°C with an average of 0.5°C in the Acropars and narrow implant group. The maximum temperature rise in T2 was 0.5°C with a mean of 0.28°C in the Duralay and narrow implant group. The least temperature rise in T2 was 0.2°C with a mean of 0.15°C, which belonged to the Duralay and wide implant group and also the Tempron and narrow implant groups. Duralay showed the highest temperature during setting (48.2°C, a mean of 45.19°C) and Acropars showed the lowest (45.2°C, a mean of 40.4°C). Tempron was in between in this regard (48°C, a mean of 40.9°C). A significant difference was found among the 3 acrylic resins in terms of temperature in T1 ($P < .001$). Bonferroni post hoc test showed that temperature rise in T1 with Duralay is significantly higher than that of the 2 other acrylic resins. The implant diameter, however, did not show any influence on the significant difference found among the 3 groups ($P = .352$), and the interaction of these 2 factors did not show any significance ($P = .422$) (Table 2; Figure 5). Two-way ANOVA showed no statistically significant difference among the different acrylic resins and the different implant diameters in T2 ($P > .05$). The interaction of these factors

TABLE 2

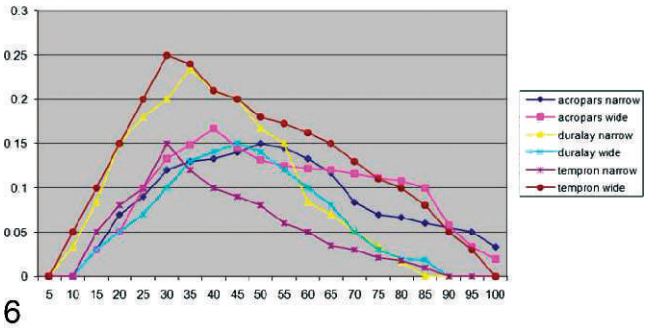
The comparative statistics of the study for the variable T1, which is the cervical point of the implant surface for which the alteration of temperature in contact with the polymerizing acrylic resin is measured

					95% Confidence Interval	
(I) Acryl	(J) Acryl	Mean Difference (I–J)	SE	*P*	Lower Bound	Upper Bound
Acropars	Duraley	−.4050*	.08690	.000	−.6197	−.1903
	Tempron	−.1800	.08690	.129	−.3947	.0347
Duraley	Acropars	.4050*	.08690	.000	.1903	.6197
	Tempron	.2250*	.08690	.037	.0103	.4397
Tempron	Acropars	.1800	.08690	.129	−.0347	.3947
	Duraley	−.2250*	.08690	.037	−.4397	−.0103

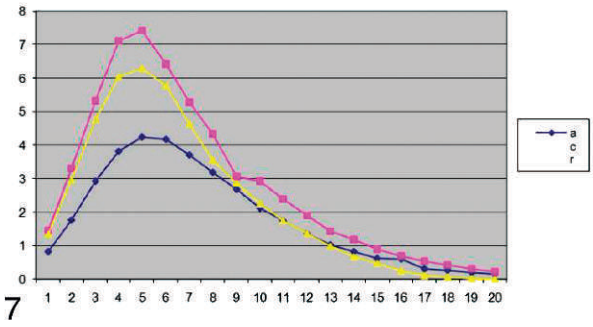
*The mean difference based on Bonferroni post hoc test is significant at the .05 level.



5



6



7

Figures 5–7. Figure 5. Comparative illustration of the temperature alterations at T1 by time according to the acrylic resin and implant diameter. X and Y axes, respectively, represent time and temperature changes. **Figure 6.** Comparative illustration of the temperature alterations at T2 by time according to the acrylic resin and implant diameter. X and Y axes, respectively, represent time and temperature changes. **Figure 7.** Comparative illustration of temperature alterations of the studied acrylic resins by time. X and Y axes represent, respectively, the time and temperature increase.

was also not statistically significant ($P = .122$) (Figure 6). Two-way ANOVA showed a statistically significant relation between the acrylic resin type and the temperature alterations recorded at T3.

Post hoc test revealed a lower temperature rise with the use of Acropars compared to the other 2 acrylic resins ($P < .001$). Duralay and Tempron acrylic resins, however, did not show any significant differences in this regard. Neither had any significance in the interaction of the 2 factors (Table 3; Figure 7).

Discussion

The present study attempted to simulate a clinical situation as much as possible. Lundskog[18] suggested the influence of circulation on the assessment of temperature alterations on the bone surface. Based on this observation, fresh sheep iliac was used to model the human jaw bone. The present study was closer to a clinical situation than the study by Ormianer et al[1] which included acrylic resin models for the insertion of implants. Also, with the application of molar-like metal molds, the amount of acrylic resin was standardized (set similar to the amount routinely applied in a clinical situation).

In the present study, the highest and the lowest temperature rise was seen, respectively, with the application of Duralay (8°C) and Acropars (4°C) (Figure 7), which is not supported by the findings of Vallittu.[2] This is thought to be attributed to the different conditions of this study. Vallittu[2] used silicon molds with different sizes, the smallest of which was a 50 × 5 × 3 mm (750 mm^3) cube. The study was performed in 22°C and 37°C. The temperature rise in the 37°C environment was 26°C, 15°C, and 12°C for the polymerized methyl, ethyl, and butyl methacrylate (PMMA, PEMA, and

Table 3

The comparative statistics of the study for the variable T3

(I) Acryl	(J) Acryl	Mean Difference (I−J)	SE	*P*	95% Confidence Interval	
					Lower Bound	Upper Bound
Acropars	Duraley	−2.5400*	.55399	.000	−3.9088	−1.1712
	Tempron	−1.7950*	.55399	.006	−3.1638	−.4262
Duraley	Acropars	2.5400*	.55399	.000	1.1712	3.9088
	Tempron	.7450	.55399	.553	−.6238	2.1138
Tempron	Acropars	1.7950*	.55399	.006	.4262	3.1638
	Duraley	−.7450	.55399	.553	−2.1138	.6238

*The mean difference based on Bonferroni post hoc test is significant at the .05 level.

PBMA) respectively. The present study shares similar conditions with the 37°C group in the study of Vallittu,[2] and the difference could be due to 2 factors. First, according to Leeson and Lippitt,[16] increased volume of the polymerizing material increases maximum setting temperature. Though the volume of the empty mold was about 400 mm^3, the placement of a solid abutment will reduce its volume of polymerizing acrylic resin to less than 250 mm^3. This volume was about 750mm^3 in the study of Vallittu,[2] which may be the reason for the discrepancy between studies. Second, the application of conductive structures like metal mold and implant has led to the establishment of an open system, which may decrease the maximum temperature during setting.

On the other hand, the mean temperature rise during the polymerization of Acropars (PBMA) compared to the other 2 groups is consistent with the findings of Brauer and Mayahra. They concluded that the maximum setting temperature depends on the type of the molecules in the chemical reaction. They reported a lower temperature peak for PBMA and PEMA compared to PMMA.

In the present study, the mean temperature rise in the cervical region was 1°C for Duralay, 0.7°C for Tempron, and 0.5°C for Acropars. Ormianer et al[1] used an intermediate volume of acrylic resin and reported the mean temperature rise in the cervical region to be 4°C to 5°C. The inconsistent findings between the 2 studies may be attributed first to the application of dense acrylic resin with a low thermal conductivity as jaw model in their study and second to the different acrylic resin volume.

The maximum thermal alterations at the apex of the fixture were recorded as 0.5°C for Duralay, 0.3°C for Tempron, and 0.3°C for Acropars. This was not consistent with the findings of Ormianer et al[1] who reported a 2°C temperature rise in this region. This could be due to the same reasons mentioned for the cervical region.

Eriksson and Albrektsson[7] studied the thermal threshold (critical temperature) beyond which thermal injury will occur. At least 7 minutes of 40°C temperature is needed for the denaturation of alkaline phosphatases within the bone. However, the present study showed that the highest temperature during the polymerization of acrylic resins and its duration is below the threshold associated with the destruction of bony cells. Therefore, despite the suggestions of Ormianer et al,[1] using cooling sprays during the acrylic resin polymerization does not seem to be helpful. In the present study, however, the standard volume of acrylic resins for relining the temporary crowns was used, and it is not clear if other proportions of powder-liquid will be as safe. According to the studies on the effect of acrylic resin volume on the maximum polymerization temperature, greater temperature rise might be expected at the bone-implant interface during the relining of the temporary removable prostheses or during the fabrication of temporary long-span fixed partial prostheses by a direct technique. This is also in accordance with the findings of the studies of Valittu[2] which recorded the maximum polymerization temperature of 3000 mm^2 PMMA to be 130°C. It is highly recommended to concern other situations like long-span fixed partial prostheses or the connection of the superstructure components through other studies. Also, the application of other acrylic resins like PMMA, which are associated with higher polymerization temperatures, should be studied further. On the other hand, it is also recommended to study the effects of polymerization temperature in different situations on the soft tissue integration of the implants.

Conclusions

Within the limitations and conditions of the present study, it may be concluded that the polymerization temperature of acrylic resins in the fabrication process of temporary single crowns does not exceed critical threshold of 47°C at the bone-implant interface. It might be further concluded that there is no risk of bone impairment associated with the direct application of acrylic resins on the placed abutments.

References

1. Ormianer Z, Laufer BZ, Nissan J, Gross M. An investigation of heat transfer to the implant-bone interface related to exothermic heat generation during setting of autopolymerizing acrylic resins applied directly to an implant abutment. *Int J Oral Maxillofac Implants*. 2000;15:837–842.

2. Vallittu PK. Peak temperatures of some prosthetic acrylates on polymerization. *J Oral Rehabil*. 1996;23:776–781.

3. Driscoll CF, Woolsey G, Ferguson WM. Comparison of exothermic release during polymerization of four materials used to fabricate interim restorations. *J Prosthet Dent*. 1991;65:504–506.

4. Shillinburg H, Hobo S, Whisett LD, Jacobi R, Bracket SE.

Fundamentals of Fixed Prosthodontics. 3rd ed. Chicago, Ill: Quintessence; 1997.

5. Wang RL, Moore BK, Goodacre CJ, Swartz ML, Andres CJ. A comparison of resins for fabricating provisional fixed restorations. *Int J Prosthodont*. 1989;2:173–184.

6. Eriksson A, Albrektsson T, Grane B, McQueen D. Thermal injury to bone. A vital microscopic description of heat effects. *Int J Oral Surg*. 1982;11:115–121.

7. Eriksson AR, Albrektsson T. Temperature threshold levels for heat-induced bone tissue injury: a vital-microscopic study in the rabbit. *J Prosthet Dent*. 1983;50:101–107.

8. Thompson HC. Effect of drilling into bone. *J Oral Surg (Chic)*. 1958;16:22–30.

9. Tetsch P. Development of raised temperature after osteotomies. *J Maxillofac Surg*. 1974;2:141–145.

10. Lavelle C, Wedgwood D. Effect of internal irrigation on frictional heat generated from bone drilling. *J Oral Surg*. 1980;38: 499–503.

11. Matthews LS, Hirsch C. Temperatures measured in human cortical bone when drilling. *J Bone Joint Surg Am*. 1972;54:297–308.

12. Mutlu G, Huggett R, Harrison A, Goodwin JW, Hughes RW. Rheology of acrylic denture-base polymers. *Dent Mater*. 1990;6: 288–293.

13. Naka S, Takeuchi T, Takahashi Y, et al. Supplemental studies on self-curing resins. 2. On polymerizing temperature of reline materials [in Japanese]. *Fukuoka Shika Daigaku Gakkai Zasshi*. 1990;17:45–52.

14. Haas SS, Brauer GM, Dickson G. A characterization of polymethylmethacrylate bone cement. *J Bone Joint Surg Am*. 1975; 57:380–391.

15. Krause WR, Park SH, Straup RA. Mechanical properties of BIS-GMA resin short glass fiber composites. *J Biomed Mater Res*. 1989;23:1195–1211.

16. Leeson MC, Lippitt SB. Thermal aspects of the use of polymethylmethacrylate in large metaphyseal defects in bone. A clinical review and laboratory study. *Clin Orthop Relat Res*. 1993; 239–245.

17. Adell R, Lekholm U, Rockler B, Branemark PIA. A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. *Int J Oral Surg*. 1981;10:387–416.

18. Lundskog J. Heat and bone tissue. An experimental investigation of the thermal properties of bone and threshold levels for thermal injury. *Scand J Plast Reconstr Surg*. 1972;9:1–80.