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In vitro infrared thermography assessment of temperature peaks during the intra-oral welding of titanium abutments

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Control of heat dissipation and transmission to the peri-implant area during intra-oral welding is very important to limit potential damage to the surrounding tissue. The aim of this *in vitro* study was to assess, by means of thermal infrared imaging, the tissue temperature peaks associated with the thermal propagation pathway through the implants, the abutments and the walls of the slot of the scaffold, generated during the welding process, in three different implant systems. An *in vitro* polyurethane mandible model was prepared with a 7.0 mm v-shape slot. Effects on the maximum temperature by a single welding procedure were studied using different power supplies and abutments. A total of 36 welding procedures were tested on three different implant systems. The lowest peak temperature along the walls of the 7.0 mm v-shaped groove (31.6 ± 2 °C) was assessed in the specimens irrigated with sterile saline solution. The highest peak temperature (42.8 ± 2 °C) was assessed in the samples with a contemporaneous power overflow and premature pincers removal. The results of our study suggest that the procedures used until now appear to be effective to avoid thermal bone injuries. The peak tissue temperature of the *in vitro* model did not surpass the threshold limits above which tissue injury could occur.

1. Introduction

The impact of various parameters on bone heating during drilling and tapping procedures used in implant site preparation and abutment preparation has been extensively studied using thermal infrared imaging. The surgery sequences and, particularly bone drilling, may lead to bone necrosis, if the tissue temperature exceeds physiological thresholds [1–4]. In particular, it is recommended that the denaturation temperature of human alkaline phosphatase (56 °C) should not be reached [5]. Recently, Olson et al. demonstrated by using infrared thermal imaging that temperatures sufficient to induce thermal necrosis of glenoid bone can be generated by glenoid preparation in shoulder arthroplasty [6]. Even after healing, uncontrolled thermal injury can result in fibrous tissue, interpositioned at the implant–bone interface, thus questioning the long term prognosis [6]. Necrosis as a result of elevated temperatures during implant site preparation has been reported in extra oral in vivo case reports [8]. *In vitro* studies in bone growth chambers have demonstrated that functional bone regeneration is achieved if the tissue temperature during the implant does not surpass 44 °C [7]. Consequently, the use of internal and external irrigation drilling systems was recommended.

In 1982 Mondani and Mondani introduced a technique for intra-oral welding of titanium implants and components in order to create a fixed restoration without the need of long laboratory procedures [9]. Hruska et al. further developed this concept and published a clinical report involving the immediate loading of 1301 implants, 436 of which were used to support partial or full-arch temporary prosthesis relined over an intraorally welded framework [10]. The failure of three of these implants (0.7%) was reported in this paper: one failed after one year of loading because its neck fractured, and two failed between the second and the third year after surgery because of periimplantitis. The authors discussed that in cases of extended reconstructions the intra-oral welding gave the advantage of simplifying the application of the fixed temporary prosthesis by overcoming the problem of disparallelism between the abutments. Furthermore, the welded framework became a mesostructure and reduced the risk of fracture or partial luting failure of the temporary prosthesis.

More than 20 years after the first paper published on the topic [9], a new protocol for the immediate loading of multiple implants by welding a pre-manufactured titanium bar to implant abutments was developed [11]. The welding procedure was conceived in order to create a customized and passive metal-reinforced provisional restoration by welding directly in the oral cavity. The titanium framework is shaped without the need of any luting agents [12] or additional interim components, such as those often necessary

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for CAD/CAM manufactured frameworks [13]. It also proved to be time and cost effective, as it employs only a simple titanium bar, commercially available denture teeth and standard titanium abutments [14]. During the welding procedure, an electrical charge from a previously loaded capacitor was transferred to the copper electrodes of the welding pincers. Electrical current supplied to the electrodes instantly raised the temperature of the two titanium components to fusion point. The process brought the core of the titanium parts to a temperature of nearly 1660 °C for 5 ms. While any discomfort was reported by the patient during the process, concerns were raised regarding heat dissipation, potential damage to the surrounding tissue, and heat transmission to the peri-implant area [14].

Therefore, the aim of this *in vitro* study was to assess the temperature peaks generated during the welding process in three different implant systems by using infrared thermal imaging. We tested both standard procedures and simulated operator's errors. To the best of our knowledge, no previous studies used infrared thermal imaging for assessing thermal effects associated with the welding procedure.

2. Materials and methods

Three implant systems were evaluated in this *in vitro* model using different combinations of abutments or welding approaches. A total of 36 welding procedures were tested on three different implant systems using six implant/abutment combinations and 1.5 mm and 2.0 mm diameter titanium bars. A standardized solid foam polyurethane mandible model was employed as scaffold for the samples. A 3.4 mm diameter 13.0 mm long XiVe® implant (DENTSPLY-Friadent, Mannheim, Germany), a 3.4 mm diameter 13.0 mm long XiVe TG® implant (DENTSPLY-Friadent, Mannheim, Germany) and a 3.5 mm diameter 14.0 mm long Ankylos® implant (DENTSPLY-Friadent, Mannheim, Germany) were used. Both standard and wrong settings were simulated in order to recreate possible errors made by the operator. The first type of error was simulated by using 100% power supply overflow instead of the standard 75–85% level recommended by the manufacturer. The second type of error was simulated by removing the copper pincers immediately after the weld, instead of keeping them in place for 3 s. In the third model, both errors were introduced. A set of tests was also conducted irrigating the titanium surfaces with physiologic saline solution at 25 °C immediately after that welding took place and using short pin screws.

2.1. Thermographic procedure

The *in vitro* sample emissivity was first estimated in order to obtain repeatable measurements. To this goal it is necessary to measure the reflected temperature of the surrounding of the specimen, which was made uniform by using an opaque black screen to avoid spotted reflections. The reflected surface temperature of the screen was measured directly by means of contact probes and its measured value passed to the data analysis software. The comparison between reflected temperature value and emitted temperature permitted to estimate the emissivity of the specimen as 0.90. After the heat distribution at depth evaluation was completed, a 7.0 mm v-shaped groove was made in the scaffold, thus making possible to visualize and evaluate the heat distribution in the 'deep' walls of the model. This allowed us to determine the surface heat distribution around the implant during the welding procedure. As the surfaces of the titanium implant, the abutments, and welding parts were excessively shiny, they were sandblasted with very hard cutting media (90 μm aluminum oxide), roughening the surface sufficiently to detect the transfer of heat and to reduce reflection. The welding clamps were also sandblasted, in order to eliminate reflection from the treated parts and rendering the entire system sufficiently emissive to be captured by means of thermal imagery. Once the sandblast procedure was terminated, the implant sample and the camera were placed at a fixed distance of 6.0 cm. All the accessories were then subsequently mounted on the selected implant and the welding procedure was carried out according to the desired parameters. Thermal measurements were performed in a climate-controlled room (temperature: 23–24 °C, relative humidity: 50 ± 5%). Thermal image series during welding procedures were obtained using a thermal camera (Series A, FLIR Systems Inc., Boston, MA, USA) (FPA: 320 × 240 of uncooled sensor working at 60 Hz; thermal sensitivity (NETD) < 0.05 °C; standard 25° optical and 45° additional lens; accuracy ±2 °C or ±2% of reading). ThermaCAM Researcher Professional software (FLIR Systems Inc., Boston, MA, USA) via a 100 Mbit connection was used to acquire and process the radiometric images. All the welds were performed using a 150 A power flow-capable intraoral-welding unit fitted with 5.0 mm diameter copper pincers (Polardent, Studioemme s.a.s., Bologna, Italy).

The measurements were performed along all the walls of the 7.0 mm v-shaped groove in the deepest section of the implant. The software detected the highest temperature captioned in the selected section by the thermal camera during the whole procedure. All the measurements were performed after the restoring the basal thermal conditions of the specimen (Figs. 1–4).

3. Results

All the temperatures recorded during the study procedures are listed in Table 1. Temperature values were measured in the zone around the implant by means of the v-slot that revealed the first 7 ms. The lowest peak temperature (31.6 ± 2 °C) along the walls of the 7.0 mm v-shaped groove was assessed in the specimens irrigated with sterile saline solution. The highest peak temperature (42.8 ± 2 °C) was assessed in the samples with a contemporaneous power overflow and premature pincers removal. The highest temperatures were reached approximately from 15 to 80 ms after the welding impulse. Peak temperatures assessed in our study lasted for nearly 10 ms in the area and dropped as time elapsed.

4. Discussion

Temperature values were measured in the zone around the implant by means of the v-slot that revealed the first 7 ms. This allowed to clearly visualize the thermal propagation through the implants and the walls of the slot. The joints were made using established parameters normally used for the oral cavity, which varied depending on the abutment type used and the selected bar diameter. The highest temperatures assessed in the measurement area never surpassed 39 °C when the recommended power flow levels for performing welding procedure were used. Errors made by the operator were also simulated in order to recreate the worst possible situation regarding heat dissipation. The first type of error was simulated by using 100% power supply to make an abutment/bar weld, which would normally be executed using 75%/85% of the maximum power flow. The second type of error was simulated by removing the pincers immediately after the welding, thus preventing heat dissipation through the copper. In the third model, both errors were introduced. An increased periimplant temperature was assessed in those models when confronted with the recommended values specimens (42.8 ± 2 °C). This temperature was, however, still far from the level considered dangerous for bone vitality (56.0 °C).

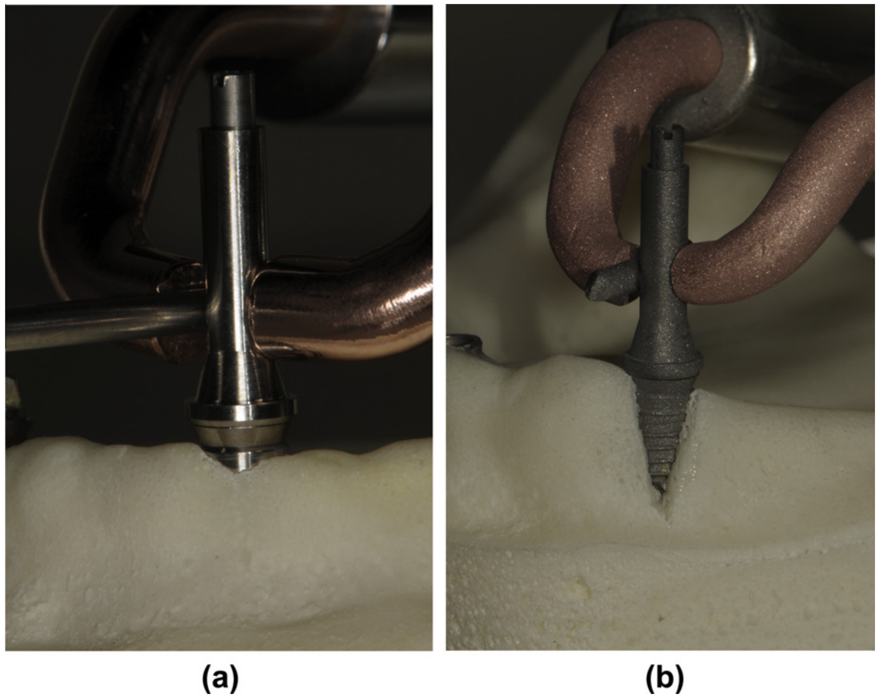

Fig. 1. Scaffold setup before (a) and after (b) the preparation of the shape.

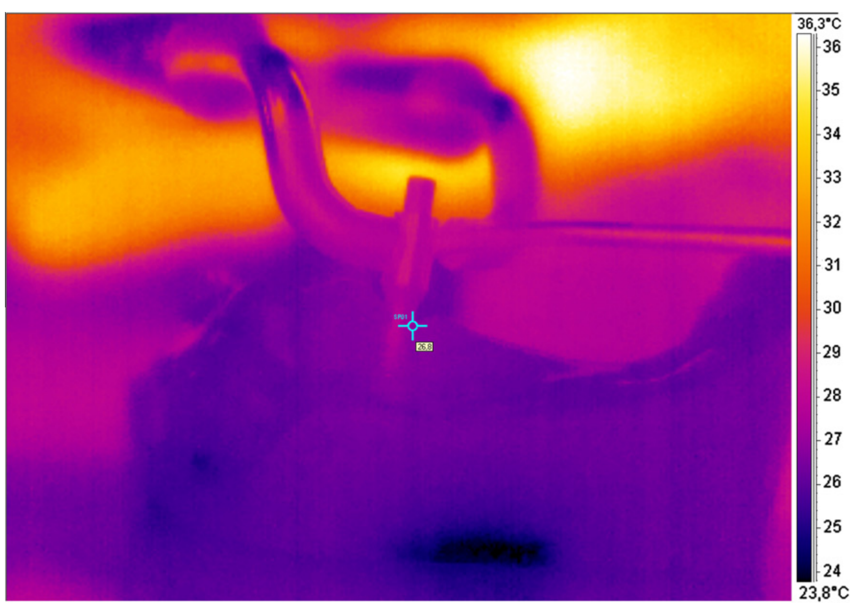

Fig. 2. Specimen 3, basal condition before welding.

It is interesting to notice that, in all specimens, peak temperature is assessed in the first implant surface/bone contact point. The highest temperatures are reached from 50 to 80 ms after the welding impulse. Peak temperatures assessed in our study lasted for nearly 10 ms in the area and dropped as time elapsed. The presence of a natural blood supply (absent in the *in vitro* specimen) may also grants better and quicker heat dissipation, with a subsequent further lowering of surface temperatures.

Heat transfer is naturally facilitated by pathways of low thermal resistance. This low resistance is determined immediately after welding by three factors: the abutment/bar joint which serves as a thermal bridge, the presence of the welding clamps and the use of cooling liquid immediately after welding.

With regard to the first factor, there was great interest in whether the presence of previously welded bars and other implants could be play a role in heat distribution and dissipation. The minor heat pathway, and thus the least favorable, is the first abutment/bar joint that is the site of the first weld. Because of this, the tests focused solely on the analysis of single welds and not on larger structure/network that may dissipate heat. As far as the second factor is concerned, it was confirmed that the presence of the clamps for at least 3 s after the completion of welding played an

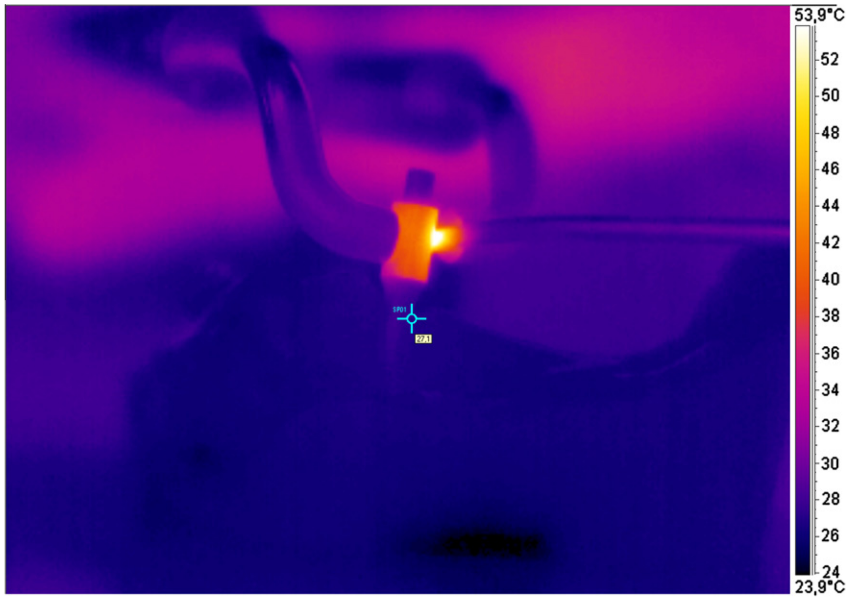

Fig. 3. Specimen 3, welding impulse, approximately 30 ms.

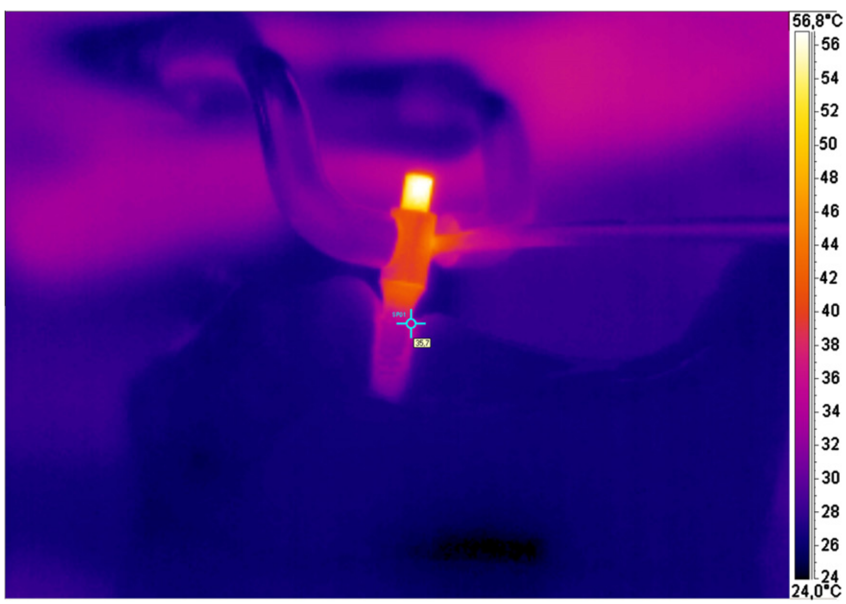

Fig. 4. Specimen 3, peri-implant peak temperature, approximately 80 ms.

important role in heat dissipation, due to its copper jaws which are excellent heat conductors and quickly disperse heat away from the framework. This study demonstrated that maintaining or removing the clamps changed heat levels around the welded point. Regarding the third factor, a set of tests was conducted irrigating the surfaces with physiologic saline solution at 25 °C immediately after welding took place. The results of our study proved that this procedure instantly interrupt the temperature rising and effectively cools the implant surface. The use of long instead of short pin screws is to be preferred. This study evidenced that standard long pin screws represented a lengthening of the thermal pathway and thus represented an alternative mean for maintaining lower temperature levels in implant areas, near vital tissue during the welding process. Further tests were also carried out by using abutments that do not require a screw attached counterpart, but rather an isodromic connection; our results demonstrated that this type of design allowed the creation of a weld with the generation of even lower temperatures than those recorded for screw type connections. This result may be explained by the increased thickness of the titanium used in the manufacture of this type of abutment. This factor, combined with the low thermal conductivity of this material in respect to the copper of the clamps, appears to create a barrier against the heat propagation in the lower zones. Thermal infrared imaging proved to be a non-replaceable technique for assessing thermal effects associated with the welding procedures. Further studies are needed, but this study poses the basis for setting up infrared imaging based real-time evaluation of the thermal effects associated with welding procedure for safer and more effective treatment. In addition, it may serve to improve the training of the specialists and trainees.

Table 1
Measurement results; Abutments: Balance Base Narrow: BBN; Syncone: SYN; Multi Purpose: MP; Temp Base: TB; TG Temp Base: TG TB. Welding components: Titanium Copings: TCO; Titanium Cap: TCA; Passive Fit: PF; Temp Base: TB; TG Temp Base: TG TB.

Implant specimen	Abutment type	Welding component	Bar diameter (mm)	Power flow (%)	Welding procedure	Peak temperature (°C)
1. Ankylos	BBN	TCO	1.5	75	Standard	35.4
2. Ankylos	BBN	TCO	1.5	75	Short pin screw	36.7
3. Ankylos	BBN	TCO	2.0	85	Standard	35.7
4. Ankylos	BBN	TCO	2.0	85	Short pin screw	36.8
5. Ankylos	BBN	TCO	2.0	85	Physiologic solution cooling	32.1
6. Ankylos	BBN	TCO	2.0	100	Power overflow	38.3
7. Ankylos	BBN	TCO	2.0	85	Premature pincers removal	39.1
8. Ankylos	BBN	TCO	2.0	100	Power overflow and premature pincers removal	40.4
9. Ankylos	SYN	TCA	1.5	75	Standard	34.9
10. Ankylos	SYN	TCA	2.0	85	Standard	35.5
11. Ankylos	SYN	TCA	2.0	85	Physiologic solution cooling	31.6
12. Ankylos	SYN	TCA	2.0	100	Power overflow	36.6
13. Ankylos	SYN	TCA	2.0	85	Premature pincers removal	38.5
14. Ankylos	SYN	TCA	2.0	100	Power overflow and premature pincers removal	38.6
15. XiVe S	MP	PF	1.5	75	Standard	37.9
16. XiVe S	MP	PF	2.0	85	Standard	38.8
17. XiVe S	MP	PF	1.5	75	Short pin screw	38.7
18. XiVe S	MP	PF	2.0	85	Short pin screw	39.8
19. XiVe S	MP	PF	1.5	75	Physiologic solution cooling	33.7
20. XiVe S	MP	PF	2.0	85	Physiologic solution cooling	33.8
21. XiVe S	MP	PF	2.0	100	Power overflow	41.7
22. XiVe S	MP	PF	2.0	85	Premature pincers removal	42.3
23. XiVe S	MP	PF	2.0	100	Power overflow and premature pincers removal	42.8
24. XiVe S	TB	TB	1.5	65	Standard	38.2
25. XiVe S	TB	TB	1.5	65	Short pin screw	39.3
26. XiVe S	TB	TB	1.5	65	Physiologic solution cooling	33.6
27. XiVe TG	n/a	PF	1.5	75	Standard	36.5
28. XiVe TG	n/a	PF	2.0	85	Standard	36.8
29. XiVe TG	n/a	PF	2.0	85	Short pin screw	38.1
30. XiVe TG	n/a	PF	2.0	85	Physiologic solution cooling	33.4
31. XiVe TG	n/a	PF	2.0	85	Power overflow	37.8
32. XiVe TG	n/a	PF	2.0	85	Premature pincers removal	40.8
33. XiVe TG	n/a	PF	2.0	100	Power overflow and premature pincers removal	41.7
34. XiVe TG	TG TB	TG TB	1.5	65	Standard	37.5
35. XiVe TG	TG TB	TG TB	1.5	65	Short pin screw	38.4
36. XiVe TG	TG TB	TG TB	1.5	65	Physiologic solution cooling	32.9

5. Conclusions

The results of this *in vitro* study suggest that the procedures recommended until now appear to be effective to avoid thermal injuries in the lower bone regions. The peak temperatures assessed in the measurement area never surpassed 39 °C when the recommended power flow levels for performing the welding procedure were used. The increased temperatures, up to 42.8 ± 2 °C, assessed in the simulated multiple error models were far from the level considered dangerous for bone vitality (56.0 °C).

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