

Stress distribution in the peri-implant area of pure titanium and titanium-zirconium small implants

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Aim: In dental implant treatment, there is a demand for mechanically stronger implants. Despite the existence of several studies showing the clinical success of narrow diameter implants, most of them are based on pure titanium (cpTi) alloys. There is a few clinical evidences of the success rate of titanium-zirconium (TiZr) narrow diameter implants. The aim of this study was to evaluate the stress distribution in the peri-implant area of narrow diameter cpTi and TiZr implants under axial and oblique loads. **Methods:** Photoelastic models were produced using epoxy resin (PL2, Vishay Precision Group) from a master model. The implants (cpTi and TiZr; Straumann AG) had 3.3 mm in diameter and 12 mm in height. Loads of 100 N and 200 N were applied to the abutment at angles of 0° (axial), 10°, 20°, and 30° (oblique). A circular polariscope (Eikonai) was used under dark field white-light configuration. The isochromatic fringes were analyzed in the peri-implant region in 5 areas, using ASTM table with isochromatic fringes; cervical-mesial, cervical-distal, mid-mesial, mid-distal and apical. **Results:** In general, under axial and oblique loads, the stress in the TiZr implant was lower than in the cpTi implant. The load of 200 N produced the highest stress values in cpTi and TiZr implants. In both implants and loads, the fringes were located more in apical area at all angles evaluated. **Conclusion:** It can be concluded that for small implants, the load inclination and intensity change the pattern of stress distribution and the cpTi implant exhibited the highest peri-implant stress.

Keywords: Dental implants. Dental stress analysis. Stress, mechanical.



Introduction

Narrow diameter implants are indicated as a clinical alternative for patients with a limited alveolar ridge or limited space. The osseointegration of commercially pure titanium (cpTi) with surrounding bone emphasize its clinical success, however, its mechanical strength can be insufficient when narrow diameter implants (<3.5 mm) are used, since the diameter directly influences the fatigue strength¹. One approach to overcome this problem is strengthening the mechanical properties of the titanium by alloying it with other materials, such as zirconium (Zr). This strategy increased the elastic modulus, hardness as well as tensile and fatigue strength, maintaining the biocompatibility similar to cpTi both in laboratory^{2,3} and clinical studies^{4,5}.

The cpTi implants with a narrow diameter have a lower mechanical strength compared with titanium-zirconium (TiZr) implants. These characteristics can influence the magnitudes of stress and consequently the outcomes of peri-implant therapy⁴. The addition of more than 50% Zr to TiZr alloy increases its resistance by two and a half fold⁵. Zirconium also reduces the melting point of the alloy and associated costs⁶. Another advantage of the addition of zirconium in titanium alloys is an improvement in the corrosion resistance with the formation of a stable oxide layer on the surface of the alloy^{7,8}. In a study investigating the effect of the percentage by weight of Zr on the mechanical properties of TiZr alloy samples, it was found that the TiZr samples had higher micro-hardness values than Ti samples at all concentrations of Zr⁹.

The success rate of TiZr narrow diameter implants has been shown to be similar to that of regular diameter cpTi implants¹⁰. The TiZr alloy has been classified as non-cytotoxic material¹¹. Furthermore, TiZr alloy has a monophasic α -structure like titanium and it allows performing surface modification using the conventional sand-blasting and acid etching procedures¹². Despite the reported benefits of narrow diameter TiZr implants for use in narrow areas, particularly the upper resistance of TiZr compared with that of cpTi²⁻⁵, literature is scarce regarding the evaluation stress distribution patterns of TiZr alloy implants in surrounding bone. Thus, it was of great interest to understand if the microtextured TiZr implant surface would present similar peri-implant stress compared with the cpTi.

The aim of this study was to assess the peri-implant stress of narrow diameter (3.3 mm) cpTi and TiZr implants under load (100 N and 200 N) in axial (0°) and oblique direction centralized (at 10°, 20°, and 30°) to the long axis of the implants. The hypothesis was that there will be difference in the number of high-intensity fringes between the type of dental implants and the loads applied.

Materials and Methods

Two photoelastic models were produced using epoxy resin (PL2, Vishay Precision Group, Wendell, NC, USA) from a master model; one with a cpTi implant and the other with a TiZr implant. The cpTi implant (Institut Straumann AG, Peter Merian, Basel, Switzerland) comprised the following: nitrogen 0:05%; carbon 0:08%; hydrogen

0.015%; iron 0.50%; oxygen 0.40%; maximum waste 0.1% each; total maximum waste 0.4%, and Ti (comprising the remaining balance). The chemical composition of the TiZr implant (Institut Straumann AG, Peter Merian, Basel, Switzerland) was 85% Ti and 15% Zr. Both implants were of the bone level type, 3.3 mm in diameter and 12 mm in height, SLActive, and had a narrow connection. Healing abutments were positioned on both implants.

The circular polariscope (Eikonal, São Paulo, Brazil) was used in the dark field configuration; therefore, the polarizer and analyzer were crossed. The optical axes of the quarter wave plates were also crossed and made an angle of 45° with those of the polarizer and the analyzer. The photoelastic models were observed before each test. The circular polariscope was utilized to verify the absence of residual stress and also for recording stresses (isochromatic fringes) during rehearsals. The polariscope was adapted to a universal test machine (AG-X 50 kN, Shimadzu, Tokyo, Japan) with a 5 kN load cell for implementing centralized and oblique axial compressive load at a crosshead speed of 0.5 mm/min. Eight trials were performed with simulated forces (100 N and 200 N) and at a central axial oblique of 0°, 10°, 20°, and 30° to the long axis of the implants.

Stress was identified with the aid of a American Society for Testing and Materials (ASTM) table which lists the sequence of colors and values for the orders of isochromatic fringes in the photoelastic material observed in a circular polariscope; in a dark field configuration with white light, under progressive increasing load (black=0; gray=0.28; light yellow=0.6; orange=0.79; intense red=0.9; red-blue transition=1.0; intense blue=1.06; blue-green=1.2; green-yellow=1.38; orange=1.62; pink-red=1.81; red-green transition=2.0; green=2.33; green-yellow=2.5; red=2.67; red-green transition=3.0; green=3.1; pink=3.6; pink-green transition=4.0; green=4.13)¹³. The following peri-implant areas were evaluated: cervical-mesial (CM), cervical-distal (CD), mid-mesial (MM), mid-distal (MD) and apical (A).

Results

The results (magnitude of stress) around the cpTi and TiZr implants (incline = 0°, 10°, 20°, and 30°), which illustrate the stresses resulting from the 100 N and 200 N loads, were based on the number of high-intensity fringes in the stress patterns in the photoelastic models (Figures 1 and 2).

Table 1 shows that for the majority of the areas evaluated, the values of the isochromatic fringe order around the TiZr implant were equal to or lower than those observed in case of the cpTi implant. The differences in stress between the two implants were small, with the exception of a few regions such as the CM and CD under an axial force of 200 N. In these areas, the TiZr implant demonstrated considerably lower stresses (0.45 and 0.45, respectively) compared with the cpTi implant (1.2 in both areas).

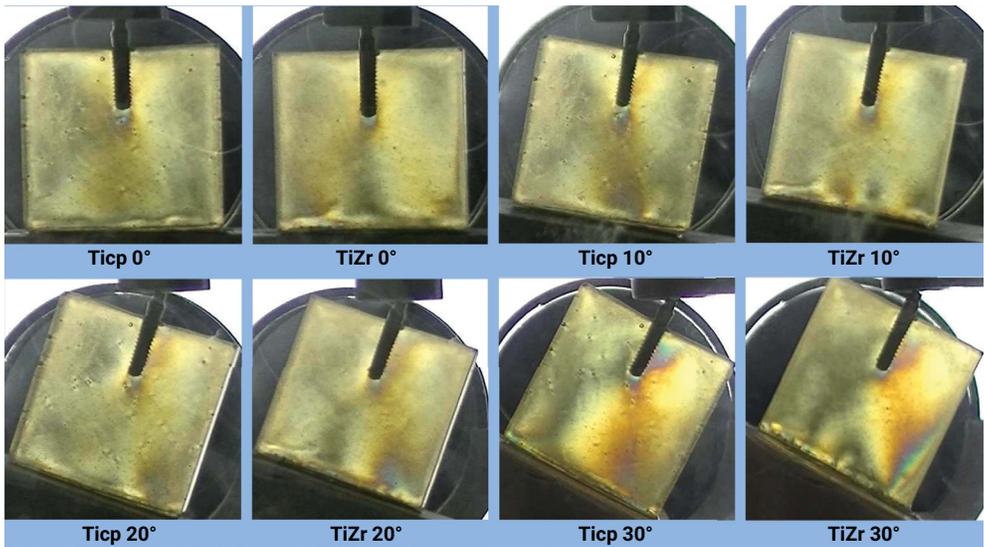


Figure 1. Photoelastic models with load of 100 N.

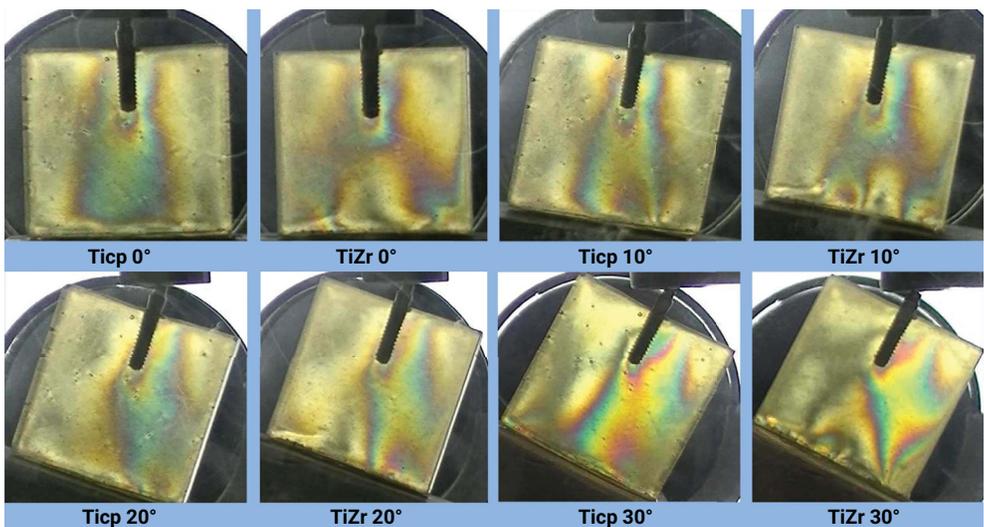


Figure 2. Photoelastic models with load of 200 N.

Discussion

The proposed hypothesis was accepted since both implants increased stress. TiZr implant showed the highest values of stresses (fringe orders). Moreover, the apical peri-implant area received the greatest stress, except when the inclination of the implant increased to 30°. In this case the highest stresses were observed in the contralateral region of the application of force (CM).

Table 1. cpTi and TiZr peri-implant stresses (fringe orders) with loads of 100 N and 200 N.

100 N								
Area	cpTi 0°	TiZr 0°	cpTi 10°	TiZr 10°	cpTi 20°	TiZr 20°	cpTi 30°	TiZr 30°
CM	0.28	0.28	0.60	0.45	0.79	0.45	1.38	1.20
MM	0.79	0.60	0.79	0.60	0.90	0.90	1.00	0.90
A	1.20	1.20	1.38	1.20	1.20	1.20	1.20	1.06
MD	0.79	0.60	0.60	0.45	0.45	0.45	0.28	0.28
CD	0.45	0.28	0.45	0.28	0.28	0.28	0.28	0.28
200 N								
Area	cpTi 0°	TiZr 0°	cpTi 10°	TiZr 10°	cpTi 20°	TiZr 20°	cpTi 30°	TiZr 30°
CM	1.20	0.45	1.38	1.38	1.81	1.81	2.50	2.33
MM	1.38	1.20	1.38	1.38	1.81	1.81	2.33	1.81
A	2.33	2.00	2.33	2.00	2.00	2.00	2.33	1.81
MD	1.38	1.20	1.06	0.60	0.60	0.60	0.45	0.45
CD	1.20	0.45	0.79	0.28	0.28	0.28	0.45	0.28

The TiZr peri-implant stresses (fringe orders) that are lower than cpTi peri-implant stresses are highlighted in bold.

Several factors have been associated with changes in peri-implant stress¹⁴. These include bone density¹⁵, the bone ridge¹⁶, type of material and intermediate prosthetic, occlusal relationship, implant connections^{16,17} and the length and diameter of the implant¹⁸. The implants investigated in this study have characteristics that in favor of the reduction of stress. Implants are intraosseous because they have lower stresses than those placed at the gingival level¹⁹, and their internal hexagon produces lower stresses due to its geometry and connection stability²⁰.

The TiZr implant demonstrated superior tensile strength (953 MPa)²¹ and fatigue strength (230 N)²² compared with the cpTi implant (680 MPa²¹ and 205 N²², respectively). This is because the modulus of elasticity of the TiZr implant (100 GPa) is smaller than that of the cpTi implant (110 GPa)¹⁹. The difference in the modulus of elasticity between the TiZr and cpTi implants has been attributed to minor stresses in TiZr peri-implants observed using finite elements¹⁹ and strain gage²³. This also supports the results of the present study, which identified lower stresses in the TiZr implant through photoelasticity. The alloys used in implants have to combine high mechanical strength with low modulus, and should be located close to the bone at 30 to 40 GPa²⁴ in order to avoid stress shielding and subsequent bone resorption. Therefore, the fact that the TiZr implant has a lower modulus of elasticity than the cpTi implant might also explain the highest quality of bone observed around this implant when compared with the cpTi implant²⁵ and the minimal bone resorption within the first two years after implantation²⁶. Although the TiZr implant has a lower elastic modulus than the cpTi implant, it is still very different from that of bone. Therefore, the future development of superior alloys requires those with similar characteristics to the alloys in this present study, but with a closer proximity to the bone's modulus of elasticity.

Qualitative analysis of the stresses indicated differences in stress between peri-implants of the same geometry in the majority of the analyzed areas. In 80% of the cases, at 0° and 10°, the stress was lower in the TiZr implant than in the cpTi implant, independent of the applied force. These cases exemplify situations similar to the angulation of the teeth in individuals with an average of 9° (mesiodistal) to the upper lateral incisors and 3° slope (buccolingual), according to Andrews' classification. The four lower incisors have an angulation (mesiodistal) average of 2° and a tilt (buccolingual) average of $-1^{\circ 27}$. The lower stress observed with the use of the TiZr implant in this study is favorable because the lower stress in the peri-implant bone reduces the potential for bone resorption and increases the commitment of the short- or long-term implant. The implant inclination in these regions should not exceed the specified value, however, in many cases implants are inserted at greater slopes to compensate for differences in bone sagittal maxillomandibular relationships.

In the present study, the highest stress levels were recorded in the apical region, except at an angle of 30°. In this case the stress was higher in the CM region. When both implants were oriented at 10°, 20°, and 30°, stresses tended to increase gradually with increasing load, and were concentrated in the contralateral side (CM and MM) of the applied load. However, on the opposite side (CD and MD), the application of load resulted in a gradual decrease in the stress under the same conditions.

The findings of previous photoelastic analysis studies of peri-implant stresses applying oblique forces are consistent with the present findings and demonstrate that the higher stresses generated are due to oblique loading²⁷. In the previous studies, the slopes varied accordingly: 0° and 10°¹, 0° and 20°²⁸, 0° and 30°²⁹, and 0° and 45°²⁷.

The masticatory load of each edentulous region under consideration for rehabilitation should be considered when selecting the implant diameter. The bite force in the region of the incisors is 14 to 25 kgf and varies according to gender and age¹. The present study utilized forces of 100 N and 200 N applied to the loads and these forces were consistent with those used in previous photoelastic analysis studies^{1,27-29}. There is a few peri-implant stress analysis studies in literature that investigated narrow diameter cpTi and TiZr implants. While this present study results supports recently published outcomes from *in vitro* studies, clinical studies are still required to confirm these findings.

In conclusion, compared with the cpTi implant, the TiZr (15% Zr) implant is associated with lower stress in the majority of peri-implant regions when subjected to a variety of loads and angles.

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