Rotational Load Fatigue Performance of Titanium vs Titanium-Zirconium Implant-Abutment Connections

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Purpose: Titanium-zirconium (Ti-Zr) alloy has been developed to strengthen the implant body, but clinically relevant information is still limited. The aim of this in vitro study was to compare the rotational load fatigue performance of implant-abutment connections in narrow-diameter (3.3-mm) and regular-diameter (4.1-mm) implants made with commercially pure grade 4 titanium alloy (CPTi-G4) and Ti-Zr. Materials and Methods: Narrow-diameter (N) and regular-diameter (R) implants with CPTi-G4 (Ti) or Ti-Zr (Tz) materials were tested. This resulted in four test groups: NTi, NTz, RTi and RTz. Five specimens were made for each group (n = 5). Abutments used were milled from titanium-aluminum-niobium alloy abutment blanks. A rotational load fatigue machine applied a sinusoidally varying load at an angle of 45 degrees to produce an effective bending moment of 35 Ncm at a frequency of 14 Hz in air at 20°C. The number of cycles to failure was recorded. The upper limit was set as 5 million cycles. Results were evaluated using analysis of variance (ANOVA) and Tukey post hoc tests. Failure locations and patterns were evaluated with scanning electron microscope (SEM). Results: All regular-diameter test groups reached the upper limit of 5 million cycles without failure. All narrow-diameter test groups failed within the range of 402,530 cycles to 3,374,353 cycles. It could be observed that NTz showed a higher mean cycle count as compared to NTi. NTi test group recorded two implants damaged, one implant fracture, five abutment fractures, and four screw fractures. NTz test group showed only abutment fractures at the level of implant platform, with no damage to the implant bodies. Significant difference was found between implants of different diameters. There was no significant difference between implants of different materials. Conclusion: Regular-diameter implants performed significantly better than narrow-diameter implants, regardless of material, while no significant difference in cyclic load to failure was found between groups of different alloys. All NTz failures were at the abutment only, without damage to the implant. This failure pattern can potentially be clinically advantageous in terms of retrieval and subsequent replacement of a failed prosthesis. Int J Oral Maxillofac Implants 2022;37:740–747. doi: 10.11607/jomi.9260

Keywords: implant fracture, implant-abutment connection, load fatigue, narrow diameter implant, titanium-zirconium

The use of dental implants for the replacement of missing teeth has been documented to be a reliable form of treatment. The reported high survival rates are largely based on studies that utilized regular-diameter implants (RDIs). However, alveolar bone dimensions are often inadequate for ideal implant placement. Narrow-diameter implants (NDIs) were introduced in the 1980s, in a bid to minimize the need for costly and time-consuming surgical augmentation. Klein et al divided NDIs into three categories: (1) one-piece with diameter < 3.0 mm (mini-implants); (2) two-piece with diameter of 3.00 to 3.25 mm; and (3) two-piece with diameter of 3.30 to 3.50 mm. Systematic reviews report good survival rates up to a mean follow-up of 2.5 to 3 years, thus concluding that NDIs can be a treatment alternative to RDIs. However, longer-term performance data and success rates for NDIs are still unknown. The incidence of implant fracture has been reported to be < 1%. However, implant fracture constitutes a catastrophic failure and requires surgical removal of the implant, with its accompanying morbidity to the patient. In addition, Shemtov-Yona and Rittel analyzed implants that experienced late biologic failure under scanning electron microscope (SEM) and found 60% of such implants to have crack-like defects or full cracks. Despite the relatively low reported incidence, the potential for implant fracture requires further investigation.

Commercially pure titanium or titanium alloys have been commonly used in dental implants due to their...
Two implant diameters (3.3 or 4.1 mm) and two implant materials (CPTi-G4 or Ti-Zr) were tested. The CPTi-G4 implants used were 12.0-mm-long Straumann Bone Level implants (3.3-mm-diameter NC 021.2612 and 4.1-mm-diameter RC 021.4612, Straumann). The Ti-Zr implants used were 12.0-mm-long Straumann Roxolid Bone Level implants (3.3-mm-diameter NC 021.2512 and 4.1-mm-diameter RC 021.4512, Straumann). Table 1 provides details of the four test groups. Specimens were labeled according to the nomenclature: diameter (N or R), material (Ti or Tz), specimen number (1 to 5). For example, the first narrow-diameter Ti specimen was labeled as NTi1. Test groups had a sample size of five (n = 5).

Machined brass specimen holders were clamped at one end onto a custom rotational load fatigue machine. The other end had a 6.0-mm-diameter cylindrical hole bored to hold the implant specimen with PL-2 resin (Vishay Precision Group). PL-2 was selected because its modulus of elasticity (2.1 × 108 N/m2) closely simulated that of natural trabecular bone (1.4 × 108 N/m2). The resin was allowed to set for at least 24 hours at 50°C in accordance with the manufacturer’s recommendation. This PL-2 resin has previously been used as the embedding medium for implants in rotational load fatigue studies.16–18 The implant was positioned to simulate 2.0-mm bone loss from the platform level.

Custom-milled titanium abutments (Straumann Premilled Abutment Blank for CARES, TAN, NC 022.0060 or RC 022.0061) were milled to a height of 10.0 mm and diameter of 8.0 mm to ensure secure and precise attachment onto the bearing housing via matching precision-milled receptacles. The abutment with bearing housing attached was then connected to the implant and torqued to 35 Ncm according to the manufacturer’s recommendations. A torque gauge (Model BTG60CN-S, Tohnichi Manufacturing) was used to ensure accurate and consistent torque levels.

The rotational load fatigue machine used in this study was similar to previous studies15–18 (Fig 1) and applied fluctuating cyclic stress onto the test specimen. The rotation of the specimen along its long axis allowed the stress at the implant-abutment connection interface to fluctuate from tensile to compressive in a sinusoidal pattern around the specimen. The test specimen was secured at 45 degrees with respect to the

### MATERIALS AND METHODS

Two implant diameters (3.3 or 4.1 mm) and two implant materials (CPTi-G4 or Ti-Zr) were tested. The CPTi-G4 implant was developed was titanium-zirconium (Ti-Zr). Zirconium is in the same periodic group as titanium and has similar chemical properties. When compared to CPTi, Ti-Zr alloy was found to have a biologic response comparable to that of CPTi,12 which makes it attractive as a material for NDIs. However, clinically relevant information for Ti-Zr implants is still limited. Altuna et al13 reported mean survival and success rates between Ti-Zr and CPTi NDIs after 1 year. Although these studies show promising results, they do not provide enough evidence of the medium- to long-term clinical and mechanical performance of Ti-Zr NDIs. It is still not known whether Ti-Zr NDIs are strong enough for application in similar clinical indications as RDIs.

The purpose of this study, therefore, was to compare the rotational load fatigue performance of implant-abutment connections in 3.3-mm- and 4.1-mm-diameter dental implants made with commercially pure grade 4 titanium alloy (CPTi-G4) and zirconium-titanium alloy (Ti-Zr).
horizontal, resulting in the load being applied on the implant-abutment interface at that angle (Fig 2). The angle of 45 degrees was chosen to create a force vector that has both a lateral and a “seating” component. The brass specimen holder, with the abutment attached, was mounted onto the rotational load fatigue machine, and concentricity was checked using a dial gauge at a threshold of 50 µm. Once confirmed, the ball bearing was cemented onto the abutment using zinc phosphate cement (Super Cement, Shofu). The test load was then attached to the ball bearing via a metal chain attached to a load basket. The test load was calculated to apply a fixed 35-Ncm bending moment at the implant-abutment interface. The test load was kept consistent with previous studies\textsuperscript{15–18} to allow for comparison.

The test environment was controlled at 20°C, and the machine was set to run at approximately 14 Hz,\textsuperscript{1–4,15–18} compliant with guidelines for non-corrosion fatigue testing of implant specimens. The cut-off cycle count was preselected to be $5 \times 10^6$ cycles in accordance with FDA specifications.\textsuperscript{19} The order of testing of the specimens was randomized across the four test groups.

Failed specimens were examined under light microscopy (SZ51, Olympus) at $\times 40$ magnification to assess failure sites and modes of failure. Representative specimens were selected for examination under SEM (Quanta 650 FEG, Thermo Fisher Scientific) for qualitative analysis.

All data were subjected to a two-way analysis of variance (ANOVA). Subsequent one-way ANOVA and Tukey highly significant difference (HSD) post hoc tests at 95% significance level were conducted. All statistical analysis was done using statistical software (SPSS version 25.0, IBM).

**RESULTS**

The results are summarized in Table 2. All RTi and RTz specimens exceeded the $5 \times 10^6$–cycle cutoff without failure.

For the NTi test group, all specimens failed, and the mean cycles to failure was 1,131,546.6 ± 568,333 cycles, ranging from 403,530 for NTi3 to 1,882,708 for NTi1. NTi1 displayed abutment and screw fractures at the level of the implant platform, without damage to the implant (Fig 3). NTi2 displayed screw, abutment, and implant fractures. The abutment fractured at the level of the implant platform, while the screw fractured at a level slightly below that. A crack on the implant was seen (Fig 4). NTi3 displayed fracture at the abutment and screw. No implant fracture was seen. On closer inspection, although the screw fractured at the implant platform level, the abutment fractured at a lower level. Damage to the anti-rotational feature of the implant can also be seen (Fig 5). NTi4 displayed fracture to the abutment and screw, with accompanying damage to the implant. The abutment failed under the level of the implant platform. Flaring of the implant platform was also seen (Fig 6). NTi5 fractured only at the abutment at the implant platform level. The implant and screw were both undamaged (Fig 7).

Similar to the NTi group, all NTz specimens failed. Cycles to failure ranged from 1,014,275 for NTz1 to 3,374,353 for NTz3, with a mean of 1,890,438.2 ± 1,124,572. All NTz specimens consistently failed at the abutment, with no damage seen on the implants. The NTz group failures were identical to the fracture mode of specimen NTi5 as representatively shown in Fig 7.

Two-way ANOVA found that the combined effect of material and diameter on cycle count was not significant ($P = .197$). The effect of diameter was found to be...
Fig 3  Fracture mode of specimen NTi1. (a) Abutment and screw fractures at level of implant platform, without damage to implant. (b) Vertical section view of NTi1. Red lines = sites of fracture. (c) SEM of failed NTi1.

Fig 4  Fracture mode of specimen NTi2. (a) Screw, abutment, and implant fractures. (b) Vertical section view of NTi2. Red lines = sites of fracture. Abutment and screw fractured near the level of the implant platform. (c) SEM of failed NTi2. Crack of implant is seen.

Fig 5  Fracture mode of specimen NTi3. (a) Abutment and screw fracture. (b) Vertical section view of NTi3. Red lines = sites of fracture. Abutment fractured below the implant platform level. (c) SEM of failed NTi3. Damage to the anti-rotational feature can be seen. (d) SEM of boxed area in Fig 5c at higher magnification, showing damage to the anti-rotational feature.
significant \((P < .001)\), while the effect of material was not \((P = .197)\).

Subsequent one-way ANOVA found significant difference between the groups \((P < .001)\). Tukey post hoc tests showed significant differences between NTi and RTi \((P < .001)\), NTi and RTz \((P < .001)\), NTz and RTi \((P < .001)\), and NTz and RTz \((P < .001)\). There was no significant difference between NTi and NTz \((P = .265)\) or between RTi and RTz \((P > .99)\).

### DISCUSSION

All test groups in this study were of the Straumann Bone Level system, with identical implant and implant-abutment connection design features, dimensions, and machining tolerances. Therefore, the results of the four test groups are directly comparable and provide insights into the material and diameter experimental variables investigated.

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**Table 2  Load Fatigue Performance of Test Groups, with Type of Failure**

<table>
<thead>
<tr>
<th>Specimen no.</th>
<th>NTi (Cycles to failure (Type of failure))</th>
<th>NTz (Cycles to failure (Type of failure))</th>
<th>RTi (Cycles to failure (Type of failure))</th>
<th>RTz (Cycles to failure (Type of failure))</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1,882,708 (A, S) 1,014,275 (A)</td>
<td>&gt; 5 million</td>
<td>&gt; 5 million</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>1,214,068 (A, S, I) 3,374,353 (A)</td>
<td>&gt; 5 million</td>
<td>&gt; 5 million</td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>403,530 (A, S, I) 2,827,987 (A)</td>
<td>&gt; 5 million</td>
<td>&gt; 5 million</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>1,383,151 (A, S, i) 1,207,259 (A, s)</td>
<td>&gt; 5 million</td>
<td>&gt; 5 million</td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>774,271 (A) 1,028,317 (A)</td>
<td>&gt; 5 million</td>
<td>&gt; 5 million</td>
<td></td>
</tr>
<tr>
<td>Mean (SD)</td>
<td>1,131,546.6 (568,333.5) 1,890,438.2 (1,124,572.1)</td>
<td>&gt; 5 million</td>
<td>&gt; 5 million</td>
<td></td>
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</table>

A = Abutment fracture; a = abutment damage; S = screw fracture; s = screw damage; I = implant fracture; i = implant damage.

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**Fig 6**  Fracture mode of specimen NTi4. (a) Abutment and screw fractures with catastrophic damage to internal connection of implant. (b) Vertical section view of NTi4. Red lines = sites of fracture. Abutment fractured below the implant platform level. (c) SEM of failed NTi4. Surface damage can be observed on all aspects of the implant connection due to loosening of abutment.

**Fig 7**  Fracture mode of specimen NTi5. (a) Abutment fracture at implant platform level, with no damage to screw head. (b) Vertical section view of NTi5. Red line = site of fracture. (c) SEM of failed NTi5. Clean fracture at abutment, with no damage to implant and screw head. This fracture mode is also representative of all NTz specimens.
Experimental Setup
The model simulated 2-mm bone loss to represent more challenging and clinically realistic conditions. The fatigue testing cutoff of $5 \times 10^6$ cycles has been speculated to be equivalent to 25 years in clinical service,\textsuperscript{20} and therefore it is more realistic to assume some degree of bone loss within this timeframe. In addition, if any of the test groups were found to survive through the cutoff limit, it is reasonable to assume that the specimens would have fared as well, or even better, if the implant had been completely embedded in bone.

In the cyclic rotational load fatigue protocol, the entire circumference of the implant-abutment interface is subjected to compressive and tensile loads, with one peak compressive and one peak tensile load every cycle. In effect, the rotational load fatigue testing protocol "seeks" out the failure location (or weakest point) of the structure. This is a major advantage of the rotational load fatigue methodology as compared to unidirectional "bend-release" fatigue testing. Rather than leaving it to chance to discover the fatigue limit of a particular orientation of the implant-abutment interface when tested, ie, peak compressive and tensile stresses at only one set of opposing locations, the rotational load fatigue protocol subjects the entire 360-degree aspect of the implant-abutment interface circumference to peak compressive and peak tensile loads to seek out failure. This could therefore be considered a more stringent and exhaustive testing protocol. Consequently, the failure outcomes observed in the failed specimens in this study are a result of fluctuating peak tensile to peak compressive stresses through the cumulative fatigue load cycles, up to the cutoff limit.

Implant Diameter
Regular-diameter implants exhibited significantly greater resistance to fatigue failure as compared to narrow-diameter implants. This can be attributed to the increased thickness of the implant-abutment complex, which has a direct impact on mechanical properties. In addition, this observation may be explained by the law of beams.\textsuperscript{21} The equation relates stress within the beam ($S$) to the force ($F$), length ($L$), and diameter ($d$) of a cylindrical specimen:

$$S = \frac{8 \times FL}{\pi d^3}$$

As the diameter of the specimen increases, resistance of the beam to bending is increased by the third power. Thus, a slight increase in the diameter of the implant-abutment complex has a large effect on the stress the material experiences.

Implant Material
Medvedev et al\textsuperscript{22} found higher endurance limits for Ti-Zr compared to CPTi-G4 when testing on dog bone-shaped specimens and on implant abutment systems. Kobayashi et al\textsuperscript{11} also found the tensile strength of titanium alloys containing 25% to 75% zirconium to be about 2.3 to 3 times greater than that of pure titanium and pure zirconium. This may explain the implant fractures found in the NTi group and lack of fractures in the NTz group.

The NTz test group showed a higher mean cycle to failure count (1,890,438.20 cycles) compared to the NTi test group (1,131,545.60 cycles), although the difference was not significant. The abutments and screws used were the same across the two groups, and so this observation could be attributed to the superior physical properties of the Ti-Zr implant. It has been reported that Ti-Zr has approximately 10% higher yield strength and 10% to 15% higher ultimate tensile strength compared to CPTi-G4.\textsuperscript{22} This translates to greater resistance to plastic deformation of the Ti-Zr implant. The implant-abutment complex was held together via screw clamping, with the screw preload being one of the critical components that contributed to its success. Adequate preload of the screw allows for the abutment, screw, and implant to function as a single entity without causing too much strain on the screw. Preload also has a shielding effect to protect the screw from fatigue damage.\textsuperscript{23} Plastic deformation of the coronal neck of the implant results in the loss of intimate contact between the components, creating spaces that result in the loss of preload on the screw. This in turn leads to earlier failure of the implant-abutment interface.

NTi test group specimens exhibited greater variability in fracture patterns, while NTz group specimens consistently fractured through the abutment only without any damage to the implant or the screw. This can again be attributed to the superior physical properties of Ti-Zr implants. Loss of compressive force of the mating surfaces brought by the preload tension of the screw allowed for greater micromovement of the abutment. This movement may be the reason for greater variability and magnitude of damage to the implant-abutment interface.

NTz had a consistent pattern of failure of only abutments fracturing at the level of the implant platform, with little to no damage to the screw head. As described earlier, with screw clamping in ideal scenarios, the implant-abutment complex will function as a single entity. With this assumption, stress exerted onto the abutment will be concentrated on the abutment at the level of the implant platform. This is corroborated by a finite element study done by Chang et al,\textsuperscript{24} which modeled the Straumann Bone Level implant. Sites of stress concentration
were the most likely sites for fatigue initiation, which explained the location of fracture for all specimens.

It was observed that only NTi specimens exhibited implant damage. These ranged from a crack in the implant of specimen NTi2 (Fig 4) to damage at the anti-rotational feature of the implant in specimen NTi3 (Fig 5) to the severe damage to the implant platform and internal features of the implant in specimen NTi4 (Fig 6). Depending on the extent and nature of implant damage, internal fit with any components will very likely be compromised, and clinical removal may be required.

Comparison with Past Studies
Quek et al,15 using the same cyclic load fatigue protocol, demonstrated a pattern of more failures with decreasing implant diameters, which was consistent with the current study. In a subsequent study, Quek et al17 tested four different regular-diameter implant-abutment systems, and failures were observed in every group. This is in contrast to the regular-diameter groups in the current study (RTi and RTz), where none of the specimens failed. This may be attributed to the implant design and manufacturing processes of each implant system. The design of an implant can affect the site and magnitude of stress concentration. Notches act as stress concentration points, and these can range from a macroscopic notch from the design of the implant, to a microscopic irregularity of the implant surface secondary to the surface treatment. This is corroborated by the results reported by Quek et al,17 where all implant fractures occurred at sites of minimum dimension or stress concentration. The manufacturing process of the material also affects the microconstituents of the metal, and the presence of inclusion particles could have an effect on fatigue resistance. These inclusions potentially affect the fatigue limit of the material by acting as nuclei for fatigue resistance. These inclusions potentially affect the fatigue limit of the material by acting as nuclei for fatigue resistance.

Seetoh et al,16 also using the same cyclic load fatigue protocol, reported abutment failure for all regular-diameter Straumann Bone Level implants with a cementable titanium abutment. The internal connection design was identical (Straumann CrossFit), so the difference could only be attributed to the material of abutments used. The metal abutments used in Seetoh et al16 were made of Ti, while the Straumann RC Premilled Abutment Blank in the current study was made with titanium-aluminum-niobium (TAN) alloy. TAN has been shown to have similar fatigue strength26 as Ti-6Al-4V alloy, which was classified as grade 5 titanium and has been known to exhibit superior mechanical properties compared to CPTi.8,9

Study Limitations
This study was conducted in a dry environment that does not simulate the oral environment. The presence of artificial saliva in an in vitro mechanical test has been reported to have a significant negative effect on fatigue performance with lower cycle counts to failure being attributed to additional corrosive attack from the saliva substitute.27 Nonetheless, the dry conditions in this study replicated the study design of similar previous studies15–18 to allow for comparison of results. In addition, the effect of a corrosive environment on fatigue performance of a material is complex. It may involve a combination of failure mechanisms, like stress corrosion cracking, hydrogen embrittlement, and corrosion fatigue.28 These confounding factors can obscure the effects of the independent variables being tested.

The results of this study can only be applied to the Straumann Bone Level system with the CrossFit design connection, and the results cannot be extrapolated to other implant systems even if the same material is used.

Clinical Implications
This study confirmed that narrow-diameter implants are significantly less fatigue resistant compared to regular-diameter implants. Regardless of the material, NDIs are at a significantly higher risk of mechanical failure. Even though short-term clinical data8,15,13,14 showed high survival rate, their indication should still be limited to anterior teeth, where forces experienced are relatively lower. Occlusion of prostheses supported by NDIs should also be controlled to limit the forces exerted onto the implant-abutment complex to ensure long-term survival.

Abutments connected to Ti-Zr implants showed slight superiority compared to CPTi implants. They failed at a higher cycle count, and thus should be at a lower risk of mechanical failure. Ti-Zr NDIs have seen high success rates in short-term clinical studies.4,29,30 However, no long-term clinical study comparing the success or survival rates between Ti-Zr NDIs and CPTi NDIs has yet been reported. Importantly, Ti-Zr specimens in this study fractured consistently at the abutment, with no damage to the implant. This allows for relatively easier retrieval and subsequent replacement of the prosthesis without further surgical intervention. As the NTi group had a consistently less unfavorable failure pattern of only abutment fracture compared to the NTi group, it would seem that when NDIs are indicated, Ti-Zr implants can be recommended over CPTi implants.

CONCLUSIONS
Within the limitations of this in vitro study on cyclic load fatigue performance of implant-abutment test groups
with different diameters and materials, the following conclusions can be drawn:

1. Regular-diameter implants performed significantly better than narrow-diameter implants, regardless of material. All narrow-diameter implant-abutment specimens failed, while no failure was observed in the regular-diameter implant-abutment groups.

2. No significant differences to the cyclic load to failure were found between groups of different alloys. However, for the narrow-diameter test groups it could be observed that NTi showed a higher mean cycle count as compared to NTi.

3. Implant fracture and damage was recorded only for the NTi group (three out of five specimens).

4. The NTi group failed only at the abutment without damage to the implant. This failure pattern can potentially be clinically advantageous in terms of retrieval and subsequent replacement of a failed prosthesis.

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