Three-Dimensional Deformation and Wear of Internal Implant-Abutment Connection: A Comparative Biomechanical Study Using Titanium and Zirconia

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Purpose: The aim of this study was to investigate the influence of the abutment material and the connection geometry on deformation and wear at the internal implant–abutment connection area (IAC), using an optical scanner. Materials and Methods: Thirty-two internal conical titanium implants, and two types of prefabricated abutments (zirconia or titanium), each (n = 8) with different connection geometries (hexagon or nonhexagon) were prepared. The inner surfaces of the implants were optically scanned before and after loading for 100,000 cycles in a simulated wet environment. The scanned data were superimposed to calculate potential three-dimensional (3D) deviations. Surfaces of the two respective implants in each group were examined using scanning electron microscopy to observe fretting wear patterns. A two-way analysis of variance (ANOVA) was used for the statistical analysis. Results: The 3D deviation (deformation) was detected at the IAC in relation to the loading direction. The average 3D positive deviation and maximum positive and negative deviations at the IAC were significantly higher with zirconia abutments than with titanium abutments, regardless of connection geometries (all P < .05). However, the average 3D negative and standard 3D deviations were similar between the two materials (both P > .05). The effect of connection geometry was not significant (P > .05). After cyclic loading, an irregular wave-pattern furrow was observed on the connection area of the implant with the titanium abutment, whereas a long and straight groove was detected on that with the zirconia abutment. Conclusion: Based on this analysis, the deformation and the wear at the IAC could be significantly affected by the material of the prefabricated abutment. Int J Oral Maxillofac Implants 2018;33:1279–1286. doi: 10.11607/jomi.6349

Keywords: deformation, implant-abutment connection, prefabricated abutment, titanium, wear, zirconia

An implant-supported dental prosthesis is composed of a dental implant, an abutment, and a retentive screw. Most abutments and implants as of today have been fabricated with commercially pure titanium or titanium alloys due to their superior mechanical properties, biocompatibility, and corrosion resistance.1 Recently, high-strength ceramics such as zirconia have been regarded as successful alternatives for esthetically demanding regions.2 Ceramic abutments have esthetic advantages over metal abutments since gingival discoloration from metal alloys can be avoided.3,4 In particular, yttria-stabilized tetragonal zirconia has many advantages, including excellent biocompatibility and high fracture toughness.5 However, zirconia may be susceptible to extensive tensile forces due to the inherent brittle nature.6

A stable implant-abutment connection (IAC) is important for long-term clinical success.7 The joint stability could be affected by the abutment material, its design, and the size of the micro-level gap at the IAC.8 The internal conical connection design at the implant-abutment interface was developed to reduce the incidence of mechanical failure.9 Internal connection types have higher mechanical stability and smaller microgaps than external “butt-joint” types.10–12

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The microgap of the implant-abutment interface may cause bacterial accumulation and gingival inflammation with the potential to induce peri-implant bone loss.\textsuperscript{13,14} Furthermore, changes in the amount of micropgaps between the implant and abutment during loading may result in fretting wear and scratches on the mating surface.\textsuperscript{15} The interfacial gap at the IAC for zirconia abutments can be approximately three to seven times greater than that of conventional titanium abutments.\textsuperscript{8}

To date, many researchers have investigated the surface deformation and wear of the internal connection areas in dental implants. Klotz and colleagues\textsuperscript{16} calculated the area of titanium particle transfer caused by fretting wear at the IAC. Stimmelmayr et al\textsuperscript{17} calculated three-dimensional (3D) changes in the implant neck region by microcomputed tomography (micro-CT) image superimposition. There have also been studies on surface deformation and wear by measuring changes in gap size using synchrotron high-resolution radiography, synchrotron micro-tomography,\textsuperscript{18} and non-destructive phase contrast x-ray microtomography.\textsuperscript{19} In the medical field, there has been a report on the amount of fretting wear of titanium hip joint implants using scanning electron microscopy (SEM) and stylus profilometer.\textsuperscript{20} However, the procedure (profilometry with a stylus) is challenging and time consuming and limited for observation, especially for the internal conical area. Recently, 3D optical scanners have been widely used in designing and fabricating dental restorations, appliances, and laboratory models.\textsuperscript{21} The scanner has a cone-like field of view and collects information about the surface and depth of the object. A point cloud is constructed using data sampled from the surface geometry. Multiple individual images are taken from designated areas. Complete representation of an object is obtained through aligning and stitching the multiple scanned images together. Possible 3D deviations between objects are evaluated by superimposition of the virtual images via a best-fit algorithm. Such 3D scanners have also been used to evaluate the clinical viability of restorative materials by measuring the degree of surface wear on opposing natural teeth.\textsuperscript{22} Since digital scanning systems allow a 360-degree view of the internal connection area without damaging the specimen, it is a suitable choice for the quantitative evaluation of deformation and wear at the implant connection area.

The purpose of this study was to quantitatively evaluate possible 3D deformation at the internal implant-abutment connection (IAC) area between the titanium implant and zirconia or titanium prefabricated abutments using an optical 3D scanner. The influence of connection geometry, such as hexagon and non-hexagon, was also investigated. The null hypothesis was that there is no difference in the degree of 3D deformation of IAC of titanium implants, regardless of the abutment material or the connection geometry.

**MATERIALS AND METHODS**

**Preparation of the Implant-Abutment Complex**

A total of 32 internal conical (11-degree taper) connection titanium (commercially pure titanium grade IV, ASTM F67) implants (TS III SA, 4.0 × 11.5 mm, Osstem) were used in this study. Sixteen prefabricated abutments made of titanium alloy (Ti-6Al-4V) and 3 mol% yttria-stabilized tetragonal zirconia (3Y-TZP) were prepared. Each group of titanium (T) abutments (TS transfer, 4.5 × 7.0 mm, gingival height 5.0 mm, Osstem) and zirconia (Z) abutments (Ziocera, 4.5 × 7.0 mm, gingival height: 5.0 mm, Osstem) had two different geometry subgroups of hexagon (-H) and non-hexagon (-NH). Each of these groups (n = 8) was categorized as T-H, T-NH, Z-H, and Z-NH with different abutment materials and types of connection geometry.

**3D Scanning: Baseline**

Prior to the experiment, each implant was scanned to serve as a reference using a high-performance 3D scanner (StereoScan R8, Breuckmann) with a miniaturized projection technique with a two-camera system. According to manufacturer’s specifications, the camera resolution of the scanner is 2 × 3,296 × 2,472 pixels, and the projection resolution is 6,144 × 4,675 pixels. The measuring depths are 180 mm for the outer camera and 36 mm for the inner camera. The defined accuracy and precision are 8 µm in accordance with the VDI 2634 guidelines. The inner conical connection and platform areas of the implants were coated with propan-2-ol-butan-propane spray to avoid light reflectance, and fixed to a full 360-degree rotating inspection jig. The IAC of each implant was scanned multi-directionally, and each scanned image was obtained and exported to a Standard Tessellation Language (STL) file.

**Dynamic Loading in Wet Environment of Implant-Abutment Complex**

Abutments were connected to the implants with corresponding abutment screws. Screws were torqued to the manufacturer’s recommended torque level (30 Ncm) using a digital torque gauge (MGT100, Mark-10) and retightened after a 10-minute interval to minimize possible relaxation embedment.\textsuperscript{23} Each implant-abutment complex was then mounted into a customized stainless steel testing jig, 3 mm below the platform level, and fixed to the loading platform of the fatigue testing machine (Model 370.02 Bionix Servo...
Hydraulic test system, MTS Systems) at 30 degrees to the vertical axis to simulate oblique loading in clinical conditions (Fig 1). A hemispherical stainless steel cap was placed onto the implant abutment to mimic the crown and to prevent deformation of the abutment from the impact rod. Each prepared specimen was immersed in sterile saline of 37°C ± 2°C (Bionix EnviroBath 10 Liter, MTS Systems), and cyclic loading was applied using a fatigue testing machine with a load precision of ± 0.5% and a sinusoidal force of 15 to 150 N at a frequency of 2 Hz. All experiments were performed in a laboratory room with temperature at 20°C ± 5°C with 30% humidity conditions, following fatigue testing guidelines of metallic materials (ISO 1099). The fixed load of 150 N in this experiment was within the physiologic clinical range.12 The entire process of dynamic cyclic loading was monitored and recorded with a strain analysis program (MultiPurpose TestWare Software, MTS Systems). Each complex was continuously loaded for each predetermined cycle without unscrewing or retightening the abutment. All prepared implant-abutment specimens underwent a mechanical loading in a simulated wet environment for 100,000 cycles.

**3D Scanning: After Loading**

After dynamic cyclic loading, specimens were untightened using a torque device. The disassembled implants were carefully cleansed in an ultrasonic ethanol bath (Saehan Cleaner, Saehan) for 10 minutes and dried with compressed air. The disassembled implants were coated and fixed to an inspection jig in the same manner as mentioned in the reference scanning protocol. Optical scanning of the internal tapered connection area of the implants was performed with a 3D scanner (StereoScan R8, Breuckmann) from multiple angles. Subsequently, the scanned images of the entire inner surface of the implants after the cyclic loading were obtained and exported to a STL file.

**Superimposition Analysis**

Quantitative measurement of the deformation at the IAC was conducted using 3D inspection software (Geomagic Control, 3D Systems). The two STL datasets were superimposed by a repeated best-fit algorithm based on the selected areas. To ensure a precise measurement, each superimposed 3D image was trimmed identically as follows: the upper trimming margin was set at 0.05 mm below the level of the implant platform; the lower margin was at approximately 3.7 mm below the upper trimming margin. This corresponds to the endpoint of the internal conical taper of the implant and removes areas beyond the field of interest. Datasets of the outer implant, including threads and beveled platform margin, were excluded from the analysis. Any 3D deviations from reference datasets were illustrated in a color-coded gradient. Positive volumetric deviations were displayed as yellow to red, while negative deviations were displayed as blue to purple. The maximum critical deviation value was set at +50 µm, and the minimum value was at −50 µm in the software. The maximum nominal deviation value was set at +10 µm and the minimum value at −10 µm, so that possible fabrication tolerances of the implant system components and surface roughness due to the scanning powder were excluded from the superimposition analysis. Hence, the exhibited deviations in this study described the possible deformations of the IAC upon cyclic loading. Average 3D deviations (positive and negative), standard 3D deviations, and maximum upper and lower 3D deviations were recorded for all implant-abutment complex groups.

**Microscopic Examination**

Two implants were selected before loading to examine the intact IAC using field emission scanning electron microscopy (FE-SEM) with a 15-kV accelerating voltage at ×250 magnification (S4700, Hitachi). After loading, all the specimens were meticulously cleansed in an ultrasonic ethanol bath for 10 minutes and then air-dried. Two respective implants from each group were then randomly chosen and examined by FE-SEM at ×30, ×250, and ×1,000 magnification (S4700, Hitachi). To examine the conical tapered region, the specimen mount from FE-SEM was tilted 35 degrees from the horizontal plane.

**Statistical Analysis**

Mean values and standard deviations of the average 3D deviations (positive and negative), standard 3D deviations, as well as maximum upper and lower 3D deviations were calculated. All outcome values satisfied the assumption of normal distribution based on the Shapiro-Wilk test. A two-way ANOVA was performed.
to determine the effects of two variables of abutment material and connection geometry and their interactions on the measured 3D volumetric deviations of implant-abutment complexes. Statistical software (IBM SPSS Version 23.0, IBM) was used. The level of statistical significance was set at .05.

**RESULTS**

All implant-abutment complexes survived the loading cycles. No implant fracture or abutment screw loosening was observed in any of the specimens. No deformation exceeding the set nominal deviation (±10 µm) presented in green was observed in the superimposed IAC of the unloaded intact titanium implant. After loading, the 3D analysis revealed the surface deviations from the reference data at the IAC of loaded complexes with nonuniform color distribution. The deviation at the IAC was detected in relation to the loading direction. The pivot side of the IAC of the loaded implant-abutment complex showed positive (yellow to red) deviations, while the opposite side showed negative (blue) deviations (Fig 2a). On the pivot side, the scanned surface at the IAC of the loaded implant-abutment complex mainly exhibited deviation beyond the reference surface of the unloaded one. On the opposite side, however, the surface of the loaded implant-abutment complex exhibited deviation away from the reference surface of the unloaded one (Fig 2b).

Mean values and standard deviations of 3D deviations of loaded IACs of four different groups are presented in Table 1. Based on the two-way ANOVA, there were no significant associations between evaluated factors: abutment material and connection geometry (all P > .05). The effect of abutment material (titanium vs zirconia) was statistically significant for average 3D positive deviation (P = .017), maximum upper 3D deviation (P = .035), and maximum lower 3D deviation (P = .021) of IACs, showing significantly higher values in groups with zirconia abutments than those with titanium abutments. Average negative 3D deviation (P = .211) and standard 3D deviation (P = .379) were not significantly affected by the material types of the prefabricated abutment. Connection geometry (hexagon vs nonhexagon) showed no statistically significant influence on the aforementioned 3D deviations of IACs in tested groups (all P > .05). Statistical differences in average 3D positive deviation were found for the pivot side of the IACs between zirconia and titanium abutments after loading, while no differences were noted on the opposite side (Fig 3).

Based on SEM investigation, some machining marks possibly from the manufacturing process were clearly seen on the intaglio surface of IACs before loading (Fig 4). Differences between preloading and postloading were detectable with various types of fretting wear and scratches on the entire surface of the IAC. Microscopic evaluation revealed that some irregular, curvy wave-like patterns with coiled furrows could be seen.

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**Table 1** Means ± SD of 3D Deviations (µm) of IACs of Mechanically Loaded Titanium Implants for 100,000 Cycles in a Saline Environment

<table>
<thead>
<tr>
<th>Groups</th>
<th>Average 3D positive deviation</th>
<th>Average 3D negative deviation</th>
<th>Standard 3D deviation</th>
<th>Maximum upper 3D deviation</th>
<th>Maximum lower 3D deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>T-H</td>
<td>15.00 ± 3.46</td>
<td>22.50 ± 4.14</td>
<td>22.00 ± 3.70</td>
<td>55.63 ± 7.80</td>
<td>110.88 ± 32.02</td>
</tr>
<tr>
<td>T-NH</td>
<td>14.25 ± 2.31</td>
<td>21.63 ± 2.00</td>
<td>20.88 ± 2.53</td>
<td>55.00 ± 8.62</td>
<td>123.50 ± 21.94</td>
</tr>
<tr>
<td>Z-H</td>
<td>16.88 ± 2.10</td>
<td>22.00 ± 2.51</td>
<td>23.00 ± 2.33</td>
<td>62.63 ± 5.32</td>
<td>132.63 ± 14.82</td>
</tr>
<tr>
<td>Z-NH</td>
<td>17.00 ± 2.20</td>
<td>19.50 ± 2.51</td>
<td>21.63 ± 2.26</td>
<td>58.88 ± 5.46</td>
<td>137.75 ± 26.59</td>
</tr>
</tbody>
</table>

**Fig 2** 3D deviation at the loaded IAC. (a) Top view of the color deviation map of the IAC after cyclic loading, showing positive (yellow to red) deviation at the pivot side and negative (blue) deviation at the opposite side. (b) Sectional schematic drawing of IAC before (black solid line) and after (color dotted line) cyclic loading, showing positive deviation at the pivot side and negative deviation at the opposite side.

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T-H = connection area of implant with titanium abutment with hexagon; T-NH = connection area of implant with titanium abutment with nonhexagon; Z-H = connection area of implant with zirconia abutment with hexagon; Z-NH = connection area of implant with zirconia abutment with nonhexagon.
on IACs coupled with titanium abutments (Fig 4). On the contrary, distinctive sharp and long grooves in a straight line were observed on IACs connected to zirconia abutments (Fig 5).

**DISCUSSION**

The results of this study showed that all implants showed plastic deformation and fretting wear at the IAC after fatigue-induced loading for 100,000 cycles. Since the average number of cycles for the mastication of humans is 250,000 per year, the 100,000 cycles of in vitro loading tested in this study may correspond to approximately 5 to 6 months of in vivo function of implant-abutment complexes. Statistically, groups with zirconia abutments showed significantly higher average positive deviation, maximum upper deviation, and maximum lower deviation of the IAC after loading than those with titanium abutments. Therefore, the null hypothesis was rejected. Deformation of titanium implants at the internal connection area obviously occurred after mechanical cyclic loading, with both zirconia and titanium abutments. However, zirconia abutments can aggravate the degree of deformation and fretting wear at the IAC of the titanium implants.

There were significant differences in positive deviations of the IACs with zirconia abutments compared to those with titanium abutments, while differences were not significant for negative deviation. Such results may be due to the loading direction, as well as the different toughness of zirconia and titanium alloys. Although
all loaded implants showed deformation after loading, direction of applied force may be particularly important, as it causes different amounts of deformation depending on where and how much load is applied. Assuming that the implants used in this study were simple nonthreaded tapered cylinders with external 4-degree taper and internal conical 11-degree taper and underwent cyclic loading with a value of “F” at 30 degrees to the vertical axis, the calculated load applied perpendicular to the pivot side (F × sin 45 degrees) of the internal tapered connection area is more than twice that for the opposite side (F × sin 15 degrees). It is speculated that the amount of load applied in this study may not be enough to make a difference on the opposite side.

Connection geometry was not statistically meaningful to deformation of the IAC in this study, which suggested a limited influence on the mechanical stability of the internal conical connection, whether the abutment had a hex structure or not.

**Fig 5** Microscopic images of IACs with different abutment materials and connection geometries after loading for 100,000 cycles. (a, d, g, j) ×30, (b, e, h, k) ×250, and (c, f, i, l) ×1,000 magnification. (a to c) Area of implant with titanium abutment with hexagon (T-H); (d to f) area of implant with titanium abutment with nonhexagon (T-NH); (g to i) area of implant with zirconia abutment with hexagon (Z-H); (j to l) area of implant with zirconia abutment with nonhexagon (Z-NH).
Several studies reported that fatigue-induced loading, such as mastication, increases, to some degree, the amount of micro-motion in the implant-abutment complex, resulting in plastic deformation and surface wear at the IAC.\textsuperscript{16–19} Cyclic loading has also proven to increase micro-motion in the implant-abutment complex.\textsuperscript{15,19,25} Synchrotron-base radiographs,\textsuperscript{15,18,26} SEM, and phase contrast x-ray microtomography\textsuperscript{15,19} studies on plastic deformation showed increases in micro-level gaps accompanied with signs of wear and wear debris on the implant-abutment mating zone after cyclic loading. In the present study, interfacial deformation of the implant was found to be variably distributed along the loading direction. On the pivot side of the loaded titanium implant, which showed positive 3D deviation, the inner conical tapered surface can be affected to a larger degree by zirconia abutments than titanium abutments. Theoretically, most of the external kinetic force applied to the abutment may be converted into the force that deforms the implant-abutment interface and the force that wears the surface of the engaged materials at the IAC, possibly due to micro-motion at the implant-abutment complex. The commercially pure grade IV titanium implants in contact with titanium alloy abutments may have shown a minor degree of wear because the materials have similar elastic moduli. Conversely, titanium implants in contact with zirconia abutments may have shown more fretting wear at the interface due to a difference in elastic moduli between titanium and zirconia, which could lead to greater 3D deviations at the IAC. In accordance with the findings of this research, previous studies reported that the amount of interfacial wear at the IAC was different between titanium and zirconia abutments, showing the higher wear tendency of zirconia.\textsuperscript{16,17} However, since there have been few studies on the mechanism of wear and deformation in the IAC interface, future studies with large sample sizes are required.

The present study was performed in a warm saline environment to simulate the oral cavity. In a previous study, the area of worn titanium powder debris was microscopically measured in order to quantitatively analyze the fretting wear throughout the inner surface of the titanium implant.\textsuperscript{16} Other studies calculated dimensional changes in areas with specific geometric shapes at the connection interface.\textsuperscript{17} However, such methods could not be applied in the present study because the titanium powder debris would be washed out in the aqueous environment, resulting in measurement errors. Therefore, the present study used a highly accurate 3D optical scanner and superimposition analysis to observe a full 360 degrees of the inner connection area of the loaded implant without damaging the specimen. However, the limitation of this method is that it was not possible to quantify the actual volume of wear in the deformed area in this in vitro study. Furthermore, differences in the implant system, IAC type, and testing regimen (number of cycles, loading angle, and loading force) of mechanical loading may affect the outcome of this experiment and require further evaluation. In addition, the accuracy of the 3D data acquisition may also be affected by the angulation and inclination of the object.\textsuperscript{22}

CONCLUSIONS

Within the limitations of this study, plastic deformation and fretting wear were evident at the IAC after cyclic loading. Zirconia prefabricated abutments showed statistically higher deformation and fretting wear at the IAC of the loaded titanium implant than titanium alloy abutments. Proper selection of abutment material may be important for the structural integrity of the IAC. Use of titanium abutments or hybrid abutments with a titanium base is recommended to minimize possible deformation and surface wear of the IAC, considering possible off-axis dynamic loading on the implant during mastication.

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