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Fracture loads and failure modes of customized and non-customized zirconia abutments

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ABSTRACT

Objective. This study aimed to evaluate the fracture load and pattern of customized and non-customized zirconia abutments with Morse-taper connection.

Methods. 18 implants were divided into 3 groups according to the abutments used: Zr — with non-customized zirconia abutments; Zrc — with customized zirconia abutments; and Ti — with titanium abutments. To test their load capacity, a universal test machine with a 500-kgf load cell and a 0.5-mm/min speed were used. After, one implant-abutment assembly from each group was analyzed by Scanning Electron Microscopy (SEM). For fractographic analysis, the specimens were transversely sectioned above the threads of the abutment screw in order to examine their fracture surfaces using SEM.

Results. A significant difference was noted between the groups ($Zr = 573.7 \pm 11.66$ N, $Zrc = 768.0 \pm 8.72$ N and $Ti = 659.1 \pm 7.70$ N). Also, the zirconia abutments fractured while the titanium abutments deformed plastically. Zrc presented fracture loads significantly higher than Zr ($p = 0.009$). All the zirconia abutments fractured below the implant platform, starting from the area of contact between the abutment and implant and propagating to the internal surface of the abutment. All the zirconia abutments presented complete cleavage in the mechanical test. Fractography detected differences in the position and pattern of fracture between the two groups with zirconia abutments, probably because of the different diameters in the transmucosal region.

Significance. Customization of zirconia abutments did not affect their fracture loads, which were comparable to that of titanium and much higher than the maximum physiological limit for the anterior region of the maxilla.

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1. Introduction

The use of zirconia abutments for dental implants in oral rehabilitation is increasing because of their superior biocompatibility and aesthetics. When a metallic abutment is used and the gingival tissue is thin [1], thus permitting light transmission [2,3], the gingival tissue would appear dark gray due to the underlying metallic abutment. This can be avoided by using ceramic abutments, which will help to achieve gingival appearance similar to that with natural teeth, especially in the anterior region [4].

The crystalline structure of zirconia changes with temperature. A monoclinic crystalline structure exists between room temperature and 1170 °C, a tetragonal structure between 1170 and 2370 °C, and a cubic structure above 2370 °C, which remains stable up to its melting point. On cooling, transformation from the tetragonal to the monoclinic phase happens at approximately 950 °C with a volume increase (∼4.5%) that can lead to catastrophic failure [5]. Thus, zirconia is subjected to degradation through tetragonal-monoclinic transformation at low temperatures (low-temperature degradation or LTD), decreasing the material's strength and increasing the risk of failure during service [6–8]. This phenomenon is exacerbated in the presence of water [5].

To overcome the problem of LTD, some oxides (CaO, MgO, Y_2O_3 or CeO_2) are added to the composition to stabilize tetragonal polycrystalline zirconia (TPZ), inhibiting the tetragonal-monoclinic transformation that induces residual stress. Thus, cracking of the material is prevented and its fracture toughness is maintained [6,8–11].

Clinically, prefabricated zirconia abutments are being used, but their emergence profiles (morphological outlines) or the space for the crown are sometimes inadequate, often requiring customization either by subtraction or addition of material [12]. Some studies have suggested that customization by reduction, followed by polishing or sandblasting, can affect zirconia's microstructures, introducing residual stresses which may influence the mechanical performance of these abutments [13].

The mechanical behavior of zirconia and titanium abutments without customization was compared by measuring their load capacity using quasi-static loading under a 30° angulation [14,15]. A study that evaluated the failure under bending of abutments with different internal connections (cone and square, cone and hexagon, cone and octagon) reported that titanium abutments presented higher bending moments at failure than zirconia ones. In addition, the presence of horizontal misfit increased the stability of zirconia abutments, demonstrating the importance of abutment design and the type of connection to the stability of the implant-abutment system [14]. Zirconia abutments prepared by CAD/CAM or manual machining and restored with disilicate crowns were subjected to dynamic and quasi-static loadings [15]. No negative effect on the fracture loads of these units was observed. On the other hand, Alqahtani and Flinton [16] found that preparation of the abutment had a negative effect on the fracture load and survival rate of zirconia abutments. They attributed the reduction in the fracture load to the reduced dimensions of the abutment or weakened mechanical properties of the zirconia material following machining. In another study considering abutments with a Morse taper connection [17], the titanium abutments failed by plastic deformation under a load higher than that required to fracture the zirconia ones.

Fractography is a well-established tool in engineering and materials science for examining and analyzing fractured surfaces [18,19]. The use of fractographic pattern and surface feature recognition has been applied in dentistry to analyze clinical ceramic restoration failure. Features such as compression curl, hackles, wake hackles, twist hackles, and arrest lines are the most commonly found markings in failed all-ceramic restorations. They allow the specific reasons for failure to be determined by identifying the failure origin and the direction of crack propagation [19–23]. Analysis of the fractured surfaces of zirconia abutments thus provides useful information about the origin of fracture; causes, direction and sequence of crack propagation; and crack interaction with microstructures. Such information is extremely useful for designing abutments to avoid future failures.

The demand for tooth-like restorations has resulted in the introduction of aesthetic, prefabricated zirconia abutments to the market. However, the clinical performance of these abutments is still unclear. Several studies [3,4,12] have evaluated these abutments with an external hexagonal connection, but little is known about the mechanical behavior of these abutments with a Morse taper connection [15–17], especially when these abutments require customization by mechanical reduction. Thus, the aim of this study was to evaluate the fracture load and fracture pattern of customized and non-customized zirconia abutments with a Morse taper connection. The null hypothesis was that abutment customization would not influence their fracture load or fracture pattern.

2. Materials and methods

For this study, eighteen Morse-taper implants of Ø4.5 mm × 11 mm (Ankylos Plus, Dentsply Friadent, Sao Paulo, Brazil) were embedded in polyurethane (F16, Axson, Cergy, France), the elastic modulus of which is similar to that of human lamellar bone tissue (Polyurethane: 3.6 GPa and lamellar bone: 4.0–4.5 GPa) [24]. The implant platforms were positioned 3 mm above the polyurethane surface to simulate 3-mm deep bone resorption [25].

Abutments for anterior teeth were considered in this study. According to the abutment used, the implant-abutment systems to be tested were divided into three groups of six:

Group Zr: zirconia abutments (Zirconia Anterior Cercon Balance, Dentsply Friadent) measuring 5.5 mm in diameter, 6.5 mm in height and 3.0 mm in gingival height;

Group Zrc: zirconia abutments (Zirconia Anterior Cercon Balance, Dentsply Friadent) measuring 7.0 mm in diameter, 7.5 mm in height and 3.0 mm in gingival height that would be customized; and

Group Ti: titanium abutments (Anterior Cercon Balance, Dentsply Friadent) measuring 5.5 mm in diameter, 6.5 mm in height and 3.0 mm in gingival height.