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Zirconia implant abutments: biomechanical behaviour

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ABSTRACT

The ceramic, aluminium oxide (alumina), was introduced in 1993, but the first fully ceramic abutment was introduced a year later, in 1994, and consisted of highly sintered alumina (CerAdapt, Nobel Biocare). However, the problem with this abutment was its fragility. The mechanical properties of zirconium oxide (zirconia) abutments were improved and they offered new opportunities for restorations. Zirconia plays a vital role in modern biotechnology because of its inertness and excellent mechanical properties of strength and hardness. This ceramic abutment is manufactured from yttria-stabilised zirconia (Y-TZP), which has been used in orthopaedic surgery for over 20 years. However, zirconia has not been used in the dentistry field for very long, so no long-term studies of its mechanical behaviour in the mouth have been conducted.

The overall objective of this work is to study the static strength and fatigue from in vitro tests on upright abutment specimen samples of the standard zirconia implant diameter made according to the standard UNE-EN ISO 14801.

The main findings of this study are as follows: all abutments break at the neck; all abutments can be used long-term in the anterior maxilla; and finally all studies on prosthetic attachments should be carried out using an established protocol (standard UNE-EN ISO 14801), to make comparisons easier between them.

KEYWORDS

Zirconia abutments; ceramic implant abutments; Breaking force; Fatigue; Zirconia abutment stress rupture.

BACKGROUND

The demand for aesthetic dental prostheses in patients is today an indisputable fact. The scientific community has spent some time researching this topic to provide solutions increasingly related with the image of a natural tooth by removing the metal and making pure ceramic prostheses.

Implant abutments have traditionally been made of metal. The use of titanium reduced galvanic and corrosive effects. Titanium abutments involve the use of metal-ceramic crowns upon them, with the aesthetic drawbacks this entails. The introduction of fully ceramic abutments improved the Vickers hardness (2000 kg/mm^2 for alumina or aluminium oxide and 1200 N/mm^2 for zirconia), the colour and design of the emergence profile meant crowns could be made with a completely ceramic coating without metal, which was more translucent. However, their fragility was still a problem under stress forces. In brittle materials, the fracture starts from a defect (e.g. a pore or crack). Forces produced from chewing, for example, can start a crack that can fracture the material. Recently, a tremendous effort has been made to improve manufacturing methods of dental ceramics and, as a result, two highly resistant ceramics have appeared on the market: made of alumina and of zirconia[1].

Alumina ceramics were introduced in 1993, but the first fully ceramic abutment was introduced a year later, in 1994, and consisted of highly sintered alumina (CerAdapt, Nobel Biocare). However, the problem with this abutment was its radiolucency and fragility[1].

Zirconium is used in dental ceramics partially stabilised with yttrium (Y-TZP). This gives exceptional qualities of hardness and bending strength, which other ceramics lack. The introduction of zirconia abutments brought improved mechanical properties and provided new opportunities for restorations[1].

Numerous researchers have studied the biomechanical properties of these abutments over the last 15 years; some of the most representative articles are listed below:

In 2001, Boudrias et al[2] indicated that ceramic abutments must only be placed in the anterior section and in premolars not subject to excessive occlusal loading, due to having a lower mechanical strength than metal. They were not considered suitable for molars, canines or incisors where there is greater than 50% overbite.

In 2001, Butz[3] compared zirconia-reinforced titanium abutments (ZiReal, 3i) with pure alumina and titanium abutments in external hexagon implants exposed to 1.2 million chewing cycles until their fracture. He found similar average fracture loads of 324N for Ti and 239 for Al.

In 2006, Att et al4 evaluated the fracture strength of zirconium dioxide implant crowns on various abutments of alumina, zirconia and titanium, which were subjected to loading and high temperature cycling. The fracture strength was 1251, 241 and 457N for the Ti, Al and Zr groups, respectively. Therefore, all abutments studied could withstand the physiological occlusal forces of the anterior sector.

González Perera[1] referred to an overall lack of long-term studies on the strength of these ceramic abutments for both single-tooth implants and for short-span bridges.

In another study in 2008, Aramouni et al[5] evaluated Certain implants and Straumann SLA ITI implants into 3 groups according to the abutments that each group had: Group 1 (Certain implants with ZiReal abutments), Group 2 (SLA implants with synOcta Ceramic Blank abutments) and Group 3 (Certain implants with UCLA noble alloy abutments). An Instron machine was used and the load applied at an angle of 45°. The fracture strength results were: Group 1 (792.7N), Group 2 (604N) and Group 3 (793.6N).

In 2011, Apicella et al6 evaluated the differences in fracture strength of titanium abutments (TiDesign 3.5/4.0) and zirconia abutments (ZirDesign 3.5/4.0, 5.5; 1.5mm). Both groups were subjected to loads until they broke. The Ti group showed significantly higher fracture strength loads (552.3 ± 23.1N), while the Zr group had a strength of 296.6 ± 45.4N. However, the authors concluded that the two types of abutments were suitable to withstand physiological mastication forces in the premolar area.

In 2013, Foong et al[7] determined the fracture strength of titanium (TiDesign, 3.5/4.0; 4.5 from Astra Tech) and zirconia abutments (ZirDesign 3.5/4.0; from Astra Tech). CAD/CAM crowns were made and a fatigue test performed at an angle of 30°. The titanium abutments fractured at an average of 270N after 81,935 cycles, while the zirconia lasted until 140N after 26,296 cycles. The fracture mode was specific for the type and design of abutment material, while the zirconia abutments fractured before the fastening screw failed.

Given the variability of results observed in the previous studies (mean fracture figures of 140N for Foong et al[7], through 296N with Apicella et al6 and up to 792N with Aramouni et al[5]), the justification of this work lies in the need to obtain sufficient and reliable scientific evidence supporting the use of zirconia abutments, while specifying the loads they are capable of supporting, for both the machined titanium base type and the entirely ceramic ones, for both internal and external connections.

The main objective of the work was to study the static and fatigue strength under load in the anterior sector via in vitro testing of a sample of straight zirconia abutment specimens with a standard diameter implant, made according to the standard UNE-EN ISO 1480188.

MATERIAL AND METHODS

The following materials were used to carry out this work:

- 6 x CAP454 abutments and 6 x Biomet 3i gold-plated screws (Biomet 3i, Palm Beach, USA).
- 6 x RC Straumann Anatomic IPS e.max straight abutments, GH 2mm, MO, O, ZrO2 and 6 titanium screws (Straumann, Basel, Switzerland).
- 6 x ZirDesign 4.5/5.0 abutments, diameter 5.5 and 1.5 mm, Astra Tech implant system and 6 titanium screws (Dentsply Implants, Mölndal, Sweden).

To perform the static testing, 9 sample holders were made according to the standard UNE-EN ISO 14801. In addition, a tool was designed for positioning the samples in the testing machine. The implants were fixed to a load-bearing, Multicore HB composite (Ivoclar Vivadent AG, Schaan, Liechtenstein). This composite was used due to its modulus of elasticity (18 GPa), which is similar to that of human bone.

The sample testing preparation method was as below:

1. Cleaning any foreign matter from inside the implants by compressed air.
2. Attaching the abutment to the implant using a screw at the different torques recommended by the manufacturer: Biomet 3i to 20 Ncm, Straumann to 35 Ncm and Astra Tech to 25 Ncm (Figure 1).
3. Fixation of a spherical attachment by adhesive to the abutment to transmit load to it. A period of at least 24 hours was left from placing the spherical attachment on the specimens until they were tested. Three samples were tested for each abutment type in the static tests and three samples per abutment for fatigue tests.

Figure 1. Abutment assembly

Figure 2. Static strength testing assembly.

Static tests were conducted using the test stand as described. The force applied induced a bending moment on the abutment as recommended by the standard UNE-EN ISO 14801. This study was conducted with a deviation from the described standard, in relation to the distance holding the sample, as the top of the implant was at the nominal bone level.

The static load tests were performed with an ELIB-20 (Ibertest, Madrid, Spain) universal testing machine at a speed of 1 mm/min using a 2kN load cell. The environmental test conditions were 20°C ± 5°C with a relative humidity of 50% RH ± 20% RH (Figure 2).

After testing was finished, the breaking force was recorded and the samples were photographed to document the failure that had occurred. The samples were stored and identified according to the study.

Fatigue testing was performed according to the standard UNE-EN ISO 14801. The installation was performed so that the load application angle was guaranteed as 28 - 32° (Figure 3).

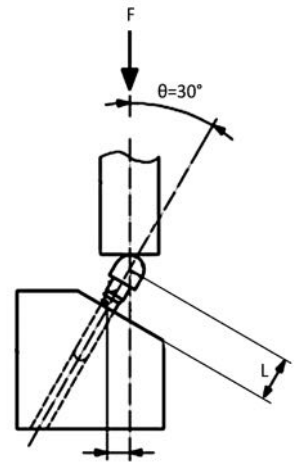

Figure 3. Testing scheme [8].

These tests were conducted with an ElectroPuls E3000 machine (Instron, Norwood, USA) at a frequency of 10Hz up to 5 million cycles, or until the abutment, screw or implant failed. The environmental test conditions were 20°C ± 5 ° C with a relative humidity of 50% RH ± 20% RH. Once the test was completed, the number of cycles was recorded and the samples photographed to document the failure that had occurred.

In accordance with the standard UNE-EN ISO 14801, the tests were performed maintaining a fatigue ratio R of 0.1 (R = Fmin/Fmax). This involves a cyclic loading oscillation during the test between a minimum value, Fmin, and a maximum value, Fmax, while keeping a constant ratio of 10%.

The Fmax value taken in each case was 25% of the aforementioned static test breaking force value.

Statistical analysis

The values obtained from the tests were expressed as the mean ± standard deviation. An analysis of variance was performed with a significance level of 5%. If there were any significant differences, a post hoc SD contrast was performed. Student's t-test was done when comparing test values before and after undergoing fatigue tests. The statistical package used to analyse the results was the SPSS 15.0 for Windows (IBM SPSS, Chicago, USA).

RESULTS

Static testing

A total of 9 abutments, 3 from each brand, were tested until breaking. Figure 4 shows the force-displacement curves for the 3 abutments studied. The highest point of the curve was taken in all cases as the breaking force point for calculation purposes.

The average breaking force was 1058 ± 225N for AstraTech, 866 ± 189N for Biomet 3i and 873 ± 402N for Straumann.

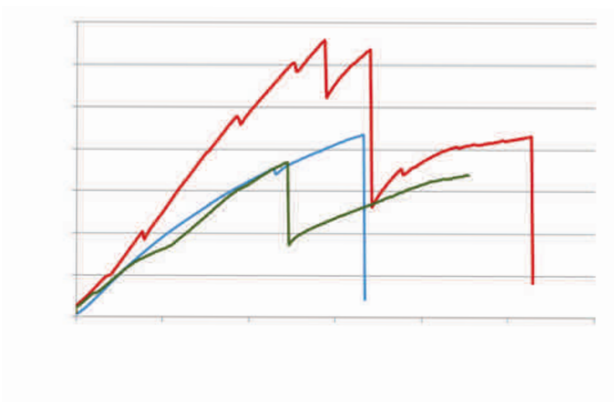

Figure 4. Force-displacement curves for the 3 abutment types studied

The inclination given to the abutment during testing (in accordance with the standard) produced a complex stress state, with peaks situated at the union of the abutment with the implant. The stress state at the abutment point with the highest load was calculated to compare the strength of the abutments of different lengths and areas, under the following assumptions:

- The abutment was considered a perfect hollow cylinder, with no peaks or protrusions.
- The stress was calculated as if the load was shared equally on the surface of the abutment upon which the load was applied.
- The maximum force was experienced at the abutment base.

This force scheme is shown in Figure 5.

Where,

F is the load applied by the testing machine.

θ is the tilt angle provided by the load block (30°).

L is the distance from the point of application of the load (F) to the support surface.

Applying a load (F) according to the standard produces a bending moment (Mf) due to the part of the load that is projected on the axis perpendicular to the abutment by the same component of the force. A constant shear force (Q) is taken into account, and a normal force (N) is produced in the direction of the implant abutment attachment point for the component in the axis parallel to the abutment.

$$M, = F \cdot Sen\theta \cdot L$$

$$N = F \cdot Cos\theta$$

$$Q = F \cdot Sen\theta$$

Where Is the angle between the direction of the load applied to the abutment (i.e. 30°). The factors required to calculate the stress on the abutment are the force (F), the distance from the load application point to the abutment (L) and the abutment cross-sectional area (A) at the point with the highest nominal load. These dimensions were determined experimentally and are shown in Table 1.

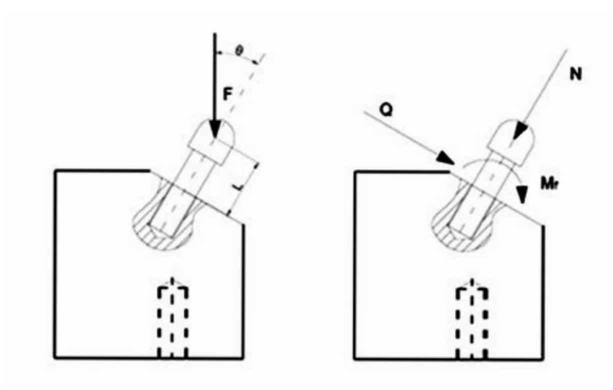

Figure 5. Distribution of forces and moments in the implant

The following procedure is followed to calculate the stresses, distinguishing between traction and compression:

$$\sigma_{trac} = -\frac{N}{A} + \frac{M_f}{I_x} \bullet r$$

$$\sigma_{comp} = -\frac{N}{A} + \frac{M_f}{I_x} \bullet r$$

$$\sigma_{1,3} = \frac{\sigma}{2} \pm \sqrt{\left(\frac{\sigma}{2}\right)^2 + \tau^2}$$

Following the Von Mises criterion:

$$\sigma_{equivalente} = \frac{1}{2}\sqrt{\left(\sigma_1 - \sigma_3\right)^2 + \sigma_1^2 + \sigma_3^2}$$

The bending moments and equivalent tensile and compression forces for each of the samples were calculated from the above expressions (Table 2).

No statistically significant differences ($p > 0.05$) were found between the stress values for the different abutments. The failure mode produced in each abutment is shown in Figure 6.

TABLE 1. ABUTMENT DIMENSIONS

	Length (mm)	Area (mm2)
Astra	8,5	9,33
Biomet	11	6,28
Straumann	11	8,16

Fatigue testing

As described in the experimental method, 25% of the static breaking force was used as the maximum test load for fatigue testing. Table 3 shows the fatigue test conditions for each of the abutments studied.

All abutments lasted for 5,000,000 cycles under these test conditions, except for one of the Biomet 3i abutments which broke at 501,497 cycles.

The abutments which survived were tested under static conditions and checked for differences before and after being subjected to the fatigue testing, to assess whether the load cycles they underwent affected their strength.

The results obtained are shown in Table 4.

The results were analysed using hypothesis testing and the Student's t-test performed at a significance level of 5%. No statistically significant differences ($p > 0.05$) were found for any of the brands between the static load values before and after subjecting them to 5,000,000 fatigue cycles at a force of 25% of the static breaking force.

DISCUSSION

The ceramic abutments with the greatest strength were made of HIP zirconia, as reflected in the numerous studies[9-11], with a tensile strength of approximately 1000MPa. However, even with these

TABLE 2. STATIC LOAD COMPRESSION TESTING RESULTS

	Force (N)	Bending) moment (Nm	Tensile stress (MPa)	Compressive stress (MPa)
Astra	1058 ± 225	4,5 ± 1,0	669 ± 142	863 ± 183
Biomet	866 ± 189	4,8 ± 1,0	894 ± 195	1131 ± 247
Straumann	873 ± 402	4,8 ± 2,2	1061 ± 488	1245 ± 573

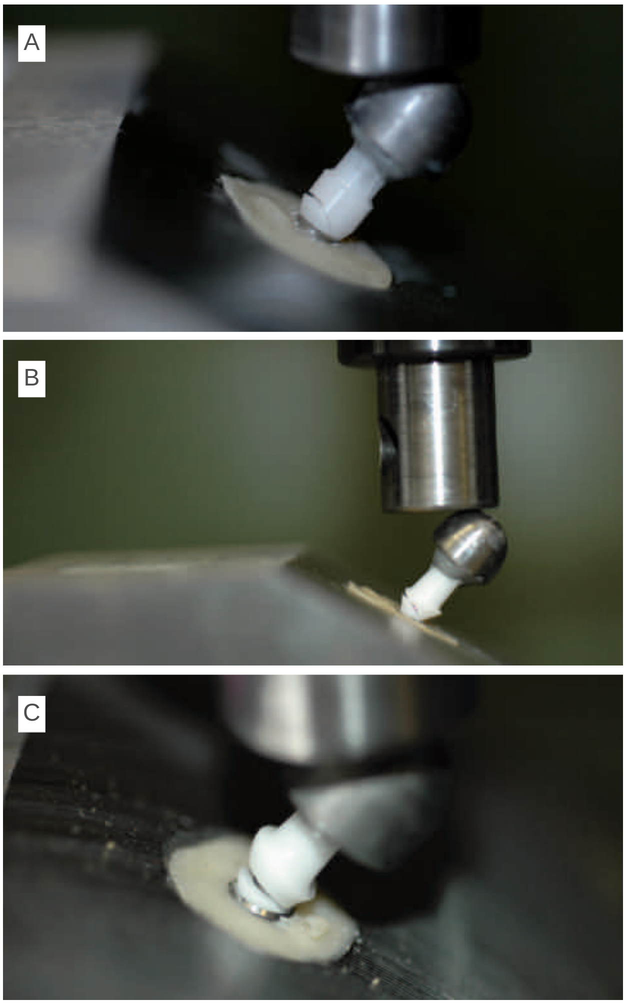

Figura 6. Fractura pilares. Biomet 3i (a), Astra (b), Straumann (c).

additions, the weak point of the "implant-ceramic abutment-screw-ceramic crown" system is the ceramic abutment. This has the greatest risk of restoration failure due to fracturing at the level of the neck. This is due to several factors: the drilling of the abutments, the abutment shape before drilling and, consequently, the stress suffered by the abutment due to the occlusal loads.

Moreover, the fracture resistance - defined as of the ability of a material to dissipate the fracture energy - of the titanium alloy used mostly in dentistry (Ti-6Al-4V) is between 84-107 MPa·m½. While the fracture resistance of zirconia (Y-TZP-HIP) is 5.5-6.7 MPa·m½. This lower fracture resistance, compared to the titanium alloy, is the major limitation of ceramic materials, as they are more susceptible to the presence of defects and so can break with giving any warning; unlike metals, which undergo plastic deformation before breaking.

To assess the strength of the zirconia abutments and their indication for use in the maxilla section where they are placed, it needs to be considered that occlusal forces in adults decrease from the molar to the incisor region; between the first and second molar, these forces vary from 400 to 800N. In premolars, canines and incisors, average forces of 300, 200, and 150N, respectively, have been recorded[12-19].

According to our study results, the abutments under static load failed at forces of 866 ± 189 to 1058 ± 225 (N), and could therefore withstand the occlusal physiological forces of the anterior sector without problems.

One of the most controversial factors in relation to the use of zirconia abutments is the observation time of the clinical studies, which include an observation of the strength of the abutments in the short term[20-24].

Exceptions are the Döring et al study[25], which had an observation period of 8 years; however, most of the abutments were made of titanium, with only 11 ceramic abutments; another study was by Ekfeldt et al[26], whose observation period was 5 years, for NobelProcera zirconia abutments made by the Biocare CAD/CAM system; and another study was by Zembic et al[27], with a 5-year observation period, which concluded that the zirconia abutments could be used very well in the posterior maxillar sectors. However, there are no clinical studies of the long-term behaviour of these abutments. Thus, in vitro or laboratory strength studies, such as this, that focus on simulating the long-term behaviour in the mouth, using fatigue tests are especially important.

The results found in reviewing the literature are very different, which may in our opinion be due to the different design of the in vitro testing.

Some studies, such as Att et al4 and Butz et al[3] differ greatly in their strength values for zirconia, alumina and titanium abutments, and not just between the two studies. Att et al[4], for example, have a large disparity between the strength values for the 3 abutment types; while Butz et al[3] have very similar strength values for the 3 types of abutment materials. Att et al[4] treated 48 maxillary central incisors on internal connection implants. Group 1 was alumina, group 2 was zirconia and the abutments control group was titanium. The crowns were cemented, and they underwent loading and high temperatures cycles. The strengths were 1251N, 457N and 241N for the Ti, Zr and Al groups, respectively. However, the Butz et al study[3] compared zirconium oxide reinforced abutments with titanium in the base (ZiReal abutment from 3i); pure alumina and pure titanium abutments in external hexagonal implants, with cemented metal crowns. They were exposed to load cycles until they fractured, with mean fracture loads of Ti (324 ± 85N), Zr (294 ± 53N) and Al (239 ± 83N). Their order of strength was consistent (Ti, Zr and Al) and all were able to withstand the physiological occlusal loads of the anterior sector; however, the results were much lower, especially for alumina. Thus, the different protocols, materials (external vs internal connection, and Zr vs Zr with titanium base) and methodology (loads and angles) in each study gave very different values. Therefore, it is considered very important to perform the tests following international standard parameters, as this has done using the UNE-EN ISO 14801 standard.

One of the most important factors that directly affect abutment performance is the design. This was seen in this study, where each abutment used was made by a different company with different dimensions and they produced different behaviour. Other authors, such as Aboushelib et al[28] and Foong et al[7], claim that the fracture mode is specific to the abutment material and design. Furthermore, other studies, such as Canullo et al[29] compare abutments from the same company, and find fewer differences between them; whereas, this study used abutments from different companies with their own designs and dimensions.

Breaking strength differences between abutments of different companies occur due to their different dimensions and designs. Thus, in our opinion, stress (MPa) and not force (Newtons) should be used to compare abutments of different dimensions and the points at which they fail.

TABLE 3. FATIGUE TESTING CONDITIONS

	Maximum force (N)	Maximum bending moment (Nm)	Tensile stress (MPa)	Compressive stress (MPa)
Astra	264,5	1,12	167,3	215,9
Biomet	216,6	1,19	223,6	282,9
Straumann	218,4	1,20	265,3	311,4

TABLE 4. STATIC LOAD TEST RESULTS AFTER FATIGUE TESTING

	Force (N)	Bending moment (Nm)	Tensile stress (MPa)	Compressive stress (MPa)
Astra	1063 ± 290	4,5 ± 1,2	672 ± 183	868 ± 237
Biomet	945 ± 61	5,2 ± 0,3	976 ± 63	1235 ± 80
Straumann	804 ± 245	4,4 ± 1,4	976 ± 298	1146 ± 350

It should also be considered that the abutments analysed in this study were tested under the conditions provided by the manufacturer, where they are usually fitted in the clinic to fit the actual situation in the mouth (the abutments were straight, with two of 11 mm length, as they usually have to be adapted to a certain inclination and a lower tooth length for the individual customer). These variations in length and inclination mean the forces the abutments can withstand vary significantly.

One example illustrating this is the Astra and Biomet abutments: the former has a shorter length and larger abutment (A) cross-sectional area, which means it can withstand a greater load before fracturing (1058 ± 225N) than the Biomet abutment (866 ± 189N), which is longer and has a smaller area.

Other authors have compared bending moments to determine the abutment behaviour[30,31]. The bending moment is produced when the force is not axial, as in our study, as anterior occlusal forces occur at an angle of 30°. The bending moment (Nm) required to fracture the abutments varied between 4.5 ± 1.0 and 4.8 ± 2.2 Nm, due to the different abutment dimensions.

Canullo et al[29] found that static testing with different abutment types gave bending moments significantly higher than those obtained by other authors, and attributed this to the dual zirconia/titanium attachment system used in these abutments.

However, as can be seen in the equations described above, the bending moment depends on the load applied and abutment dimensions; this should be considered carefully when evaluating the abutment strength.

Thus, a good assessment of abutment behaviour can be made by comparing the stress at which it breaks. The tensile stress endured by the abutments in our study ranged between 580-1612 Mpa, depending on the abutment dimensions and bending moment. If the stress that breaks the abutments is compared, it is observed that the Astra abutments failed at a stress of 669 ± 142 MPa, the Biomet at 894 ± 195 and the Straumann at 1061 ± 488, with no statistically significant differences between them ($p > 0.05$). As can be seen, the Astra abutment is the one with the least strength as it fails at the lowest stress; however, it can withstand the greatest force (1058N), if this parameter is compared.

From the dental point of view, the Astra abutment might be considered the best choice, as it can withstand a greater force (simply because it is shorter and has a larger area). However, the behaviour of the material for this abutment is the worst (as it fractures at a lower stress than the other abutments, whose fracture points are higher and closer to the theoretical tensile strength of zirconia).

Another factor not addressed in this study is the abutment design which can make the stress behaviour of the abutments vary significantly.

Table 4 shows the tensile stress is less than the compressive stress. However, the abutments failed after the crack developed on side of the abutment under tensile stress, although less than the compressive stress. The tensile stress produced is considered to be the main reason for the fracture at the abutment base, as the crack started and spread in this area.

Another important point to consider for the long-term good behaviour for restorations on a fixed prosthesis on an implant is the location of the zirconia abutment in the dental arch. There is no unanimity of criteria, however, to determine the arch position that would ensure adequate long-term clinical behaviour. In reviewing the literature, a wide range of ceramic abutments were placed in different locations on the jaw, which highlights the absence of objective, scientific evidence for positioning the zirconia abutments in the maxillary arch[20-25]. The data obtained in our work and the mean physiological occlusal forces in an adult suggest their use in the maxillary posterior area is not appropriate for long-term survival; they can be placed only in the anterior or premolar areas not subjected to excessive occlusal loading. This finding is in line with results obtained by other authors, such as Boudrias et al[2] and Cho et al[32]. According to Gehrke et al[16], it is reasonable to demand the abutments withstand up to 300N for the anterior area to 1000N for the posterior area.

This study showed a high long-term abutment survival rate, as only 1 abutment failed after being subjected to 5 million load cycles.

Comparing the static strength data after the fatigue test (Table 4) showed there were no statistically significant difference ($p > 0.05$) between the static load breakpoints before and after subjecting them to 5 million fatigue cycles, at a force of 25% of the static breaking force. These results confirm the abutments were not been damaged and maintained their initial strength.

There are several study limitations which need to be considered when making a proper correlation with clinical application. Firstly, studies with a larger number of samples are needed to obtain more representative results. Secondly, future studies should explore higher loads in fatigue testing.

CONCLUSIONS

1. Zirconia abutments fracture at the neck when overloaded.
2. The load (force) withstood by the abutment is strongly influenced by the abutment dimensions and positioning.
3. In our study, the zirconia abutment strength was unaffected by fatigue testing of 5,000,000 cycles using 25% of the static breaking force; so they have good long-term behaviour.
4. The zirconia abutments appear to be suitable for use in the anterior maxilla area.

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