An *in vitro* comparison of fractural strength of zirconia custom abutments with different angulations and thicknesses

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Abstract:

Statement of the problem: The strength of zirconia custom abutments could be affected by the thickness of the abutments' walls and/or abutments' angulations, leading to internal stress build-up, and consequently affecting restoration's life-span.

Purpose: The purpose of this *in vitro* study was to compare the fractural strength of zirconia custom abutments with different thicknesses and angulations.

Materials and methods: Eighty zirconia custom abutments were divided into two main groups (group I & group II). Each group was divided further into four subgroups. Groups I-A-1, I-B-1, II-A-1, and II-B-1 simulated a clinical situation with an ideal implant position from a prosthetic point of view, which allows for the use of a straight zirconia custom abutments with two different thicknesses (.7 mm and 1 mm). Whereas groups I-A-2, I-B-2, II-A-2, and II-B-2 Simulated a situation with a compromised implant position requiring 15 degree angulated abutments with different thicknesses

(.7 mm and 1 mm). Implant replicas were placed in cubic self cure acrylic jigs to support the abutments in all groups. The zirconia custom abutments were engaged to the implant replicas into the cubic acrylic jigs by using a manual torque wrench. Each jig was secured and mounted in metallic vice. All groups were subjected to shear stress till failure with a crosshead speed of .5 mm/min with the force transferred to the lingual surface of the zirconia custom abutments 2 mm below the incisal edge by using universal testing machine. The universal testing machine was controlled via a computer software system which also completed the stress-strain diagram and recorded the breaking fractural strength.

Results: The mean fractural strength of zirconia custom abutments in group I (A-1 through B-2) ranged from 162.49 ± 63.21 N to 231.14 ± 95.21 N, whereas in group II (A-I through B-2) ranged from 431.83 ± 96.74 N to 745.93 ± 274.79 N. Group II-B-2 exhibited the highest fractural strength among all the groups, which was statistically significant p= .005.

Conclusions:

- Zirconia custom abutments in group II exhibited the highest fractural strength especially with angulated abutments.

- Zirconia custom abutments in group I exhibited the lowest fractural strength especially the angulated abutments.

- The thickness of zirconia custom abutments in group II had a positive influence on the fractural strength, but in group I had no influence.

Dedications

To my beloved Parents,

To my family,

&

To people in new Libya

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I am so grateful to Allah (God), "Who taught *man* by the pen. Taught man what he knew not." (Qur'an), the Almighty God, for all the bounties he grants me throughout my life.

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Introduction

Dentistry has aimed to replace missing teeth since it was first recognized as a profession. For centuries, dental practitioners have relied on their own skills and various artifacts to develop esthetic and functional alternatives to minimize sequel that occurs as a result of edentulism. Partial, complete, fixed, or removable dentures are by far the most commonly used forms of tooth replacement applied. In other words, these devices have been incorporated into the oral cavity anchored on either remaining teeth and/or other anatomical structures. Only scarce archeological reports have demonstrated attempts of incorporating prosthetic devices into the jaws as more natural and functional replacements. However, predictability of these methods was not achieved until recently. [1]

During the past 40 years the porcelain-fused-to metal technique has been extensively used in fixed partial dentures (FPDs).[2] However, this technique has improved the demand for more aesthetic materials with biocompatible properties for fabricating FPDs. In addition, the public scare

about side effects of dental metals and alloys has accelerated the development of alternatives to metallic dental restoration such as different types of dental ceramics.[3–9] One of these alternatives is metal free restorations; in these types of restorations, translucency and opacity can be controlled based on patients' needs. Because of the translucency of all-ceramic restorations, it is possible to reproduce the look of natural teeth.[10]

All-ceramic single crowns are widely used both in anterior and posterior regions; these crowns have been evaluated in many follow-up studies, mostly showing promising results.[11-13] During the past decade, all-ceramic FPDs manufactured using various dental ceramics have also been evaluated in *in vitro* and in short-term *in vivo* studies.[14-19] However, dental ceramics are brittle and their low fractural resistance and relatively low flexural strength still limits the possibility of manufacturing FPDs using all-ceramic frameworks. [19]

Currently, Zirconia has been proven a biocompatible, esthetic, and functional material for long-lasting restorations. This high-strength ceramic material allows for light transmission at the critical interface between marginal gingival tissue and prosthetic components. Clinical observations have shown clearly favorable esthetic outcomes and better translucency compared to ceramo-metallic restorations.[20]

Literature Review

Historical Background:

Zircon has been known as a gem since ancient times. The name of the metal, zirconium, comes from the Arabic word, *Zargon* (golden in color), which in turn comes from the two Persian words, *Zar* (Gold) and *Gun* (Color).[21] Zirconia, the metal dioxide (ZrO_2) was identified as such in 1789 by the German chemist Martin Heinrich Klaproth in the reaction product obtained after heating the gems, and was used blended with rare earth oxides as pigment for ceramics.[21]

 The first paper concerning biomedical application of zirconia was published in 1969 by Helmer and Driskell.[22] However, the first paper concerning the use of zirconia to manufacture ball heads for Total Hip Replacements (THR), which is the current main application of this ceramic biomaterial, was introduced by Christel et al in 1988.[23] The first proposal of the use of zirconium oxide for medical purposes was made in 1969

concerned orthopedic application. ZrO_2 was proposed as a new material for hip head replacement instead of titanium or alumina prostheses.[24] Orthopedic research focused on the mechanical behavior of zirconia, on its wear, and on its integration with bone and muscle. Moreover, these first studies were largely carried out *in vivo* because *in vitro* technology was not yet sufficiently advanced.[25]

Prior to 1990, many other studies were performed, in which zirconia was tested on bone and muscle without any unfavorable results.[25-30] Since 1990, *in vitro* studies have also been performed in order to obtain information about cellular behavior towards zirconia.[31] *In vitro* evaluation confirmed that ZrO_2 is not cytotoxic.[32-34] Uncertain results were reported in relation to zirconia powders that generated an adverse response.[35-36] This was probably due to zirconium hydroxide, which is no longer present after sintering; thus, solid samples can always be regarded as safe. Mutagenicity was evaluated by Silva and Covacci, and both reported that zirconia is not

able to generate mutations of the cellular genome.[37-38] Specifically, mutant fibroblasts found on ZrO_2 were fewer than those obtained with the lowest possible oncogenic dose compatible with survival of the cells.[39] Moreover, zirconia oxide creates less flogistic reaction in tissue than other restorative materials such as titanium.[40] This result was also confirmed by a study about peri-implant soft tissue around zirconia healing caps in comparison with those around titanium.[41] Inflammatory infiltrate, microvessel density, and vascular endothelial growth factor expression were found to be higher around the titanium caps than around the ZrO_2 ones. Also, the level of bacterial products, measured with nitric oxide synthase, was higher on titanium than on zirconium oxide.[42]

In the early stages of development, several solid solutions (ZrO_2-MgO, ZrO_2-CaO, and ZrO_2-Y2O$_3$) were tested for biomedical applications.[43] Their mechanical properties are the highest ever reported for any dental ceramic. This may allow the realization of posterior fixed partial dentures and permit

a substantial reduction in core thickness. These capabilities are highly attractive in prosthetic dentistry, where strength and esthetics are paramount.[44]

Properties of Zirconia:

Zirconia is a crystalline dioxide of zirconium. Its mechanical properties are very similar to those of metals and its color is similar to tooth color.[45] In 1975, Garvie proposed a model to rationalize the good mechanical properties of zirconia and by virtue of which, it has been called ''ceramic steel''. [46]

Zirconia crystals can be organized in three different patterns: monoclinic (M), cubic (C), and tetragonal (T). By mixing ZrO_2 with other metallic oxides, such as MgO, CaO, or $Y2O_3$, greater molecular stability can be obtained.[45]

Yttrium-stabilized zirconia is presently the most studied combination.[46] The aforementioned three phases are present in a common ZrO_2 crystal. Every transition between the different crystalline reticulations is due to a force on

the zirconia surface, and this produces a volumetric change in the crystal where the stress is applied. When a stress occurs on a zirconia surface, cracking energy creates a T-M transition. This crystalline modification is followed by an expansion that seals the crack ~4-5%.[47] ZrO_2 stabilized with Y_2O_3 has better mechanical properties than other combinations; although its sintering is much more difficult, this is the principal kind of zirconia considered for current medical use. [47]

Physical Properties of Zirconia:

Zirconia's resistance to traction can be as high as 900-1200 MPa and its compression resistance is about 2000 MPa.[48-49] Cyclical stresses are also tolerated well by this material. Cales et al found that with applying an intermittent force of 28 KN to zirconia substrates, 50 billion cycles were necessary to break the samples, however, with a force in excess of 90 KN structural failures of the samples occurred after just 15 cycles. [50] Surface treatments can modify the physical properties of zirconia. Exposure to

wetness for an extended period of time can have a detrimental effect on its properties.[51] This phenomenon is known as zirconia aging. Moreover, also surface grinding can reduce toughness.[52] Kosmac et al confirmed this observation and reported a lower mean strength and reliability of zirconium oxide after grinding. [53]

Zirconia Polymorphs:

At ambient pressure, unalloyed zirconia can assume three crystallographic forms as it has been mentioned, but this formation is depending on the temperature.[54] At room temperature and upon heating up to 1170ºC, the symmetry is monoclinic. Between 1170ºC and 2370ºC, the structure is tetragonal, and 2370ºC to melting point is cubic.[55] While cooling down, the transformations from the tetragonal to the monoclinic phase occurs and they are accompanied with a substantial increase in the volume (about 4-5 percent) which is sufficient to lead to catastrophic failure.[56] The transformation is irreversible and begins at 950ºC upon cooling. In order to

prevent this transformation, a procedure of alloying pure zirconia with stabilizing oxides such as CaO, MgO, Y_2O_3, or CeO_2 allows the retention of the structure at room temperature.[57-58]

Among the three classes, the Tetragonal Zirconia Polycrystalline (TZP) materials have the best fractural toughness and mechanical strength, especially when 3 mol % yttria is used as stabilizer. Such results are related to the greater extent of yttria's solubility in tetragonal zirconia solid solution when compared to others oxides.[59]

The 3 mol% yttria-stabilized zirconia (3Y-TZP) exhibits a very important feature related to the polymorphic transformation for monoclinic phase when a mechanical stress is applied. This phenomenon, known as transformation toughening, can prevent crack growth, which results in a material with high toughness and mechanical strength.[60] At a crack tip, the matrix constraint on the tetragonal particles of 3Y-TZP is reduced by tensile stresses so that a transformation to the monoclinic structure takes place. This transformation

produces a local 4-5 percent increase in volume, which results in compressive stresses within the matrix, thereby increasing the energy necessary for further crack growth.[61-62] On the other hand, this transformation also altering the phase integrity of the material and increasing the susceptibility to aging. Aging (the low temperature degradation) of zirconia is a well-documented phenomenon exacerbated noticeably by the presence of water.[63-65] The consequences of aging are many and include surface degradation with grain pullout and micro cracking as well as strength degradation.[65]

The flexural strength of zirconia oxide (3Y-PTZ) is in the range of 900 to 1100 MPa, which is approximately twice as strong as alumina oxide ceramics, and five times greater than standard glass ceramics.[66-67] Also, the fractural strength of fully sintered 3Y-TZP may be increased further by hot isostatic pressing (HIP). Under a hot isostatic press 3Y-TZP, the material is

compressed under a pressure of 1,000 bar at 50 K below its sintering temperature.[68]

The mechanical properties of 3Y-TZP strongly depend on its grain size.[66-68] Above a critical grain size, 3Y-TZP is less stable and more susceptible to spontaneous t→m transformation, whereas smaller grain sizes (<1µm) are associated with a lower transformation rate.[69] Moreover, below a certain grain size (<0.2µm), the transformation is not possible, leading to reduced fracture toughness.[70] Consequently, the sintering conditions have a strong impact on both the stability and mechanical properties of the final product as they dictate the grain size. Higher sintering temperatures and longer sintering times lead to larger grain sizes.[71-72] Also, due to its biocompatibility; zirconia is used as a prosthetic implant for many medical and dental applications. Its chemical stability has made it an optimal material particularly in corrosive environments. Moreover, zirconia has a hard and dense surface that is ideal for resisting wear and contact damage,

which together have made zirconia the material of choice where high functional demands are to be satisfied.[60]

Fabrication of Zirconia Restorations:

3Y-TZP is available in dentistry for the fabrication of custom abutments and fixed partial dentures. Zirconia frame works cannot be manufactured by dental technicians using the traditional method of powder/liquid slurry. Frameworks have to be milled from prefabricated blanks using a digital process called Computer Aided Design/Computer Aided Manufacturing (CAD/CAM). [73]

The Accuracy of Digital Scanners in CAD/CAM:

The guidance provides the FDA's recommendations to manufacturers for evaluating and labeling optical impression systems for CAD/ CAM dental restorations. [73] An optical impression system for CAD/CAM dental restorations is a device used to record the topographical characteristics of teeth, dental impressions, or stone models by analogue or digital methods for

use in the computer assisted design and manufacturing of dental restorative prosthetic devices.[73] Such systems may consist of a camera, a scanner or equivalent type of sensor, and a computer with software.[73] Accuracy is a measure for the digitizing quality of the measured points. An existing standard for characterizing 'Digitizing quality' of coordinate measuring machines has already been devised in an international standard, but the test methods are laborious and not dedicated to the geometries and undercut measurements that are encountered in dental surface digitization.[74, 75] May et al measured the precision of fit of the crown fabricated with CAD/CAM technology for the premolar and molar teeth fit to a die and found that the mean gap dimensions for marginal openings, internal adaptation, and precision of fit for the crown groups were below 70 μm.[76] Mehl et al reported that the accuracy of digital scanners can be up to 13, 2 +/- 3.6 micrometers.[77]

Digital Data Acquisition and Virtual Design:

Computer Aided Design (CAD): CAD/CAM systems are based on the optical or mechanical acquisition of topographic data for the surfaces of specific objects. The scanning of casts is widespread and has been successfully used for many years. Impressions now are also being scanned, followed by direct intraoral data acquisition conoscopic holography.[78]

Conoscopic Holography: this is a precision scanning process introduced into dentistry to meet a wide range of requirements, capturing data at a very high level of accuracy. Like the triangulation method most dental scanner systems employ, this method works by directing a beam of laser light at the object.[74] The most important difference between the two methods is that with conoscopic holography, the source of light and the sensor is not placed at an angle, but the light reflected by the scanned object is parallelized. [78] Once the laser beam passes through the special crystal, so-called fringe patterns are created directly on the sensor.[78] The co-linear property of the

beams makes for precise measurements and robustness in the case of optical disturbances and has the added advantage of facilitating the capture of a great variety of geometric shapes such as very steep slopes or depressions.[76] There are no limiting optical shadows such as those that invariably occur with the triangulation method when attempting to capture deep cavities with a small opening radius. This makes the technology useful for scanning impressions. However, an efficient working and production process does not solely depend on how the components are fabricated.[78] Another important aspect is a software user interface that supports the operator in the virtual design task. Today, appropriate software not only helps define the most suitable shapes and sizes for crowns and bridges, but it also facilitates, based on a scanned bite record, representing the occlusal morphology and indicating and planning occlusal contacts adapted to the situation in the antagonistic jaw. In the second step, the virtual restoration is automatically cut back to reflect the future ceramic veneer, ensuring an ideal and

homogenous thickness–indispensable for long-term clinical success, especially when using frameworks made of high-strength oxides such as zirconia.[78]

Computer Aided Manufacture (CAM):

After receiving the information as digital file which has been designed by the digital scanners, the restorations are processed in the manufacture either by soft machining of pre-sintered blanks followed by sintering at high temperature, or by hard machining of fully sintered blocks.[79]

Soft Machining: This technique allows the frameworks to be ground out of zirconia in the pre-sintered, soft stage. The pre-sintered frameworks, however, have to be sintered to full density in order to reach optimal material properties. This sintering procedure is accompanied with high sintering shrinkage of zirconia of about 22 percent. In order to compensate for the shrinkage, the size of the milled, pre-sintered frameworks has to be larger by this difference.[80-83]

Typically, the 3Y-TZP powder used in the fabrication of the blanks contains

a binder that makes it suitable for pressing. The binder is eliminated during a

pre-sintering heat treatment. This step has to be controlled carefully by

manufacturers, particularly the heating rate and the pre-sintering

temperature.[81] If the heating rate is too fast, the elimination of the binder and

associated burn out products can cause the blanks to crack. Therefore, slow

heating rates are preferred. The pre-sintering temperature of the blanks

affects the hardness and machinability.[82] These two characteristics act in

different ways: an adequate hardness is needed for the handling of the

blanks, but if the hardness is too high, it might be detrimental to the

machinability. The temperature of the pre-sintering heat treatment also

affects the roughness of the machined blank.[39] Overall higher pre-sintering

temperatures lead to rougher surfaces. The choice of a proper pre-sintering

temperature is thus critical.[39] The density of each blank should carefully be

measured so that the appropriate compensating shrinkage is applied during

final sintering.[79] The final density of the pre-sintered blanks is about 40

percent of the theoretical density (6.08 g/cm^3). The density gradient within

the blanks is lower than 0.3 percent of the theoretical density in all

directions. Machining is better accomplished in two steps.[80] A first rough

machining is done at a low feed rate while the final fine machining is

performed at a higher feed rate.[79-84] Soft machining techniques prevent the

stress-induced transformation from tetragonal to monoclinic and leads to a

final surface virtually free of monoclinic phase, unless grinding adjustments

are needed or sandblasting is performed.[84] Most manufacturers of 3Y-TZP

blanks for dental applications do not recommend grinding or sandblasting to

avoid both the T $\rightarrow$ M transformation and the formation of surface flaws that

could be detrimental to the long-term performance, despite the apparent

increase in strength due to the transformation-induced compressive

stresses.[85]

Hard Machining of (Y-TZP): 3Y-TZP blocks are prepared by pre-sintering at temperatures below 1500°C to reach a density of at least 95 percent of the theoretical density. [86] The blocks are then processed by hot isostatic pressing at temperatures between 1400°C and 1500°C under high pressure in an inert gas atmosphere. This latter treatment leads to a very high density in excess of 99 percent of the theoretical density. [87]

Zirconia processing in its density sintered stage makes the grinding procedures difficult; it is time-consuming and leads to the high wear of milling instruments. Also, the restorations produced by hard machining of fully sintered 3Y-TZP blocks have been shown to contain a significant amount of monoclinic zirconia.[59] This is usually associated with surface micro cracking higher susceptibility to low temperature degradation and lower reliability.[88]

Zirconia Custom Abutments:

The good long-term clinical results of dental implants allow their applications in various regions of the arches and for divers indications.[89, 90] Following the functional and biologic success of implant supported restorations, additional criteria (primarily esthetic aspects) gain importance. Standardized titanium abutments exhibit high survival rates because of their excellent physical properties.[91] However; their application can impair the esthetic result.[92] In the case of soft tissue recession, exposure of the gray titanium abutment can lead to failure of the restoration in highly visible anterior regions. Furthermore, when titanium abutments are used in patients with a thin labial mucosa, a grayish discoloration of the mucosa can occur, owing to the gray metal color showing through it.[92, 93] The esthetic shortcomings of titanium led to the development of ceramic materials as an alternative for esthetically demanding anterior and premolar regions.[93] The first of all ceramic implant abutments were developed in 1993 and were

made of alumina, a high-strength ceramic.[94] Among the ceramics, alumina exhibits favorable physical material properties.[94,95] The bending strength of alumina reaches 547 MPa and its fractural toughness is 3.55 MPa m$^{1/2}$.[96] These early abutments were customized manually to meet the anatomical requirements of each individual site. *In vitro* studies demonstrated high fractural resistance of alumina abutments.[97] In clinical investigations, however, abutment fractures were observed in 1.9 percent of implant supported fixed dental prostheses.[98,99] In the search for ceramic abutment material with improved physical properties, yttria stabilized zirconia was introduced in 1996.[100] This high strength ceramic exhibits fractural toughness and bending strength that are almost twice as high as alumina ceramic.[96,101] The bending strength of zirconia is 900 MPa and its fractural toughness reaches 9 MPa m$^{1/2}$.[102] Zirconia abutments showed resistance to a high load of up to 730N in one *in vitro* study.[103] In comparison, the naturally occurring mean inciso-occlusal loads in anterior regions amount to 110N for

teeth and 370N for implants.[104,105] With this data from *in vitro* studies,

zirconia was expected to reduce the risk of fracture.[106] Indeed, no fracture of

zirconia abutments were reported after four years of clinical service in one

study.[106] Because of its different material properties, the mechanism of

ceramic abutment failure differs from what can be seen with titanium

abutments. Ceramics are brittle and therefore do not withstand tensile forces

very well. [103] Fracture occurs when the tensile forces exceed the limits given

by the fracture toughness. In contrast, metals are ductile. Their ductility

enhances the tolerance for compressive and tensile forces. Prior to fracture,

the first elastic deformation followed by plastic deformation occurs. This

property is the reason for the excellent loading capacity of metals. [107] The

nature and the direction of the load have a major influence on the stability of

ceramic implant abutments. During the occlusal loading of an implant

supported reconstruction, the region around the abutment screw head is the

area of the highest torque and stress concentrations, and it has been

demonstrated to be the most critical region for the stability of ceramic abutments.[108] High tensile forces occurring in the region during function were the most frequent origin of fracture of ceramic abutments in several *in vitro* studies.[103,107] With metal abutments, the same forces first led to deformation and then to fracture of the abutment screws.[109] Consequently, the type and architecture of the implant abutment connection might have a substantial influence on the stability and fixation of brittle ceramic abutments. Implants are designed with different types of implant–abutment connections.[89] The abutments can either be fixed onto an external connecting part of the implant or internally into the implant.[89] The internal connection of zirconia abutments can be accomplished either by the abutment itself (one-piece) or by means of secondary components (two-piece). One piece abutments are made entirely of ceramic, whereas for two-piece abutments, the internal connecting part can be either a secondary titanium abutment or separate metallic insert mounted on the implant together with the abutment

and fixed by means of one abutment screw.[110] The influence of the type of connection on the long term stability of the abutment-implant complex has been analyzed for titanium abutments in several studies.[111] With these types of abutments, mechanical problems such as loosening or fracturing of the abutment screw can occur with external connections.[110-112] In one clinical study of an external connection implant system, abutment screw loosening was the most frequent technical complication observed after three years of service.[111,112] In contrast, the internal connection was demonstrated to exhibit significantly higher strength *in vitro* than the external hexagonal connection, owing to its higher resistance to bending.[113] The occurrence of abutment screw fractures were lower with an internal connection.[113-115] In one clinical study analyzing an internal-connection implant system, the cumulative survival rate for the abutment screws and the restorations supported by titanium abutments was 100 percent after 18 months, and no screw loosening or fracturing occurred.[116] The stability of zirconia abutments with

external connections has been analyzed in detail.[103,117] In contrast, to date, the stability of internally connected ceramic abutments has not been specifically investigated.[117] It might be expected that one-piece and two-piece internally connected zirconia abutments exhibit different resistance to loading as a result of a different distribution of the loading forces.[117]

Zirconia Abutments' Survival Rate:

Little data is available on the survival rate and average lifetime of zirconia custom abutments.[118] The incidence of complications associated with zirconia abutments was determined by evaluating data from three studies.[118-120] One article presented the clinical success of 36 experimental zirconia abutments on single-tooth implants, after a mean observation period of 49.2 months.[118] No abutment fractures were observed during clinical loading, resulting in a cumulative survival rate of 100 percent. Abutment screw loosening was reported for two restorations (one at eight months, and one at 27 months).[118] Healthy peri-implant mucosa and stable marginal bone levels

were documented at zirconia abutments.[118] In another study, 30 zirconia abutments on single-tooth implants were observed after a follow-up period of 40 months.[119] No abutment fractures or screw loosening were reported, resulting in a cumulative survival rate of 100 percent.[119] Finally, a third study evaluated the success rate of 37 zirconia–alumina composite abutments (ZirAce, Acucera Inc., Reno, NV). Nine implants were single-tooth, and 28 implants were FPDs. After a 12-month follow-up, no abutment fractures, cracks, screw loosening, or peri-implant infection signs were reported.[120] Also, neither zirconia–alumina abutment failures nor adverse soft tissue reactions were observed at 12 months.[120] These studies indicate that zirconia abutments could be suitable for clinical use.[120]

Objectives:

The aim of this *in vitro* study was to compare the fractural strength of zirconia custom abutments with different thicknesses and angulations

Hypothesis:

There is no difference between the zirconia custom abutments with different angulations and thicknesses.

Clinical significance of the study:

The results of this *in vitro* study will help the dental practitioners with their decision making process in selecting the type of zirconia custom abutment to be used clinically.

Materials and Methods:

Experimental design: This *in vitro* study measured and compared the fractural strength of zirconia custom abutments (ZCAs) made in two different angulations (straight and 15 degree) and two different thicknesses (.7 mm and 1 mm) in two different designs one-piece ZCAs (Straumann, Minuteman Road Andover, Mass), and two-piece ZCAs (Nobel Biocare, Yorba Linda, Calif), Table 1.

This study was conducted by using 80 (n= 80) CAD/CAM (Computer Aided Design / Computer Aided Manufacturing) Zirconia Custom Abutments (ZCAs). The samples were divided into two main groups, each main group represented different implant system [Group I: Straumann system, and group II: Nobel Biocare system]. Each group was divided into two subgroups (A and B). Each subgroup had ZCAs made with two different thicknesses (.7 mm, and 1mm). These also divided further into groups based on the angulations (0^0 and 15^0), Figures 1 & 2.

Table 1: Type of zirconia abutments and their corresponding implants, connection design, and torque used for fixation of the abutments

Groups	A*	Ts**	Abutments	Implants	Type of connection	Secondary component
I-A-1	0°	.7mm	Cares RC	Straumann Bone level RC 4.1×12mm	Internal connection (cone & octagonal)	None
I-A-2	15°	.7mm	Cares RC	Straumann Bone level RC 4.1×12mm	Internal connection (cone & octagonal)	None
I-B-1	0°	1mm	Cares RC	Straumann Bone level RC 4.1×12mm	Internal connection (cone & octagonal)	None
I-B-2	15°	1mm	Cares RC	Straumann Bone level RC 4.1×12mm	Internal connection (cone & octagonal)	None
II-A-1	0°	.7mm	Procera RP	NobelReplace Select straight TiUnite RP, 4.3×13mm	Internal connection (Triangle)	Metallic Insert
II-A-2	15°	.7mm	Procera RP	NobelReplace Select straight TiUnite RP, 4.3×13mm	Internal connection (Triangle)	Metallic Insert
II-B-1	0°	1mm	Procera RP	NobelReplace Select straight TiUnite RP, 4.3×13mm	Internal connection (Triangle)	Metallic Insert
II-B-2	15°	1mm	Procera RP	NobelReplace Select straight TiUnite RP, 4.3×13mm	Internal connection (Triangle)	Metallic Insert

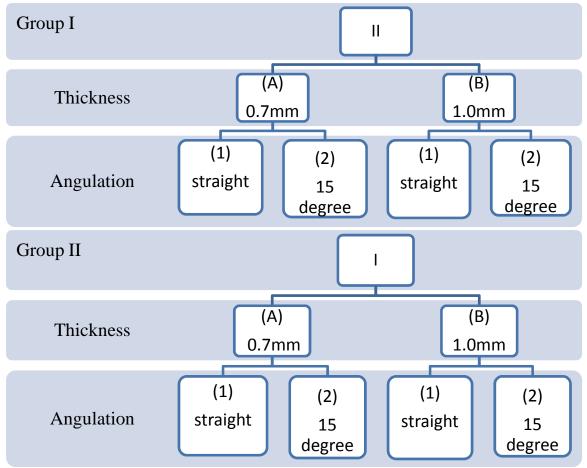RC= regular crossfit; RP= Regular plateform, A*= Angulation, Ts**= Thicknesses
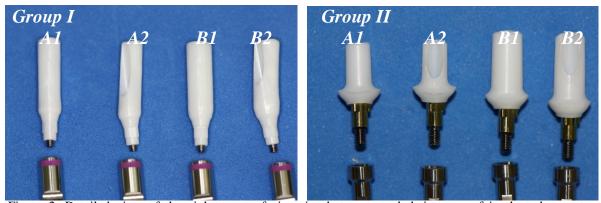
Figure 1. Groups distribution

Figure 2. Detailed view of the eight types of zirconia abutments and their types of implant-abutment connection design.

Samples Preparation:

Group I: Forty Implant replicas (n= 40) (10 for each subgroup) [Straumann

implant replica, 4.1mm×12mm   regular crossfit (RC) octagon restorative

platform, Andover, Mass] were   placed in cubic self cure acrylic jigs

[DENTSPLY Caulk, Caulk® Orthodontic Resin, York, PA].   The acrylic

jigs were standardized with dimensions 2.5 cm x 2.5 cm x 2.5 cm. To make

sure that the implant replicas were adjusted to be perpendicular to the jig's

surface (90° degrees), every replica was attached to a lab surveyor [Dentsply

Netech Yucaipa, California] by using a guide pin [Impression post

Straumann RC 4.1 mm, Andover, Mass]. A water scale was used to adjust

the implant replicas with the surveyor's pen, Figures 3, 4, 5 & 6.

Figure 3. A guide pen attached to lab surveyor.

Figure 4. Implant analog was attached with the a guide pen to be perpendicular to the base which were adjusted with a water scale.

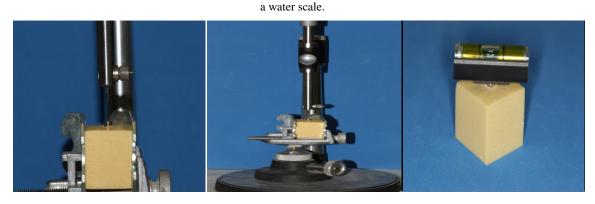Figure 5. Placing the implant analog inside the acrylic jig.

Figure 6. Acrylic jig after replacing the implant analog.

Zirconia Custom Abutments group I:

The implant replicas which were placed in cubic acrylic jigs were scanned in order to design custom abutments digitally at the manufacture facility (Straumann, Andover, Mass) by a single operator using a surface scanner (Straumann® Cares® Scan CS2, Andover, Mass), Figure 7.
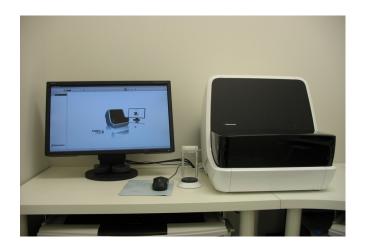
Figure 7. Straumann® Cares® Scan CS2.

This scanner uses a laser beam to trace a custom abutment and render a digitized image of the implant analog; then designs the custom abutments digitally. The finish line was set and adjusted as necessary using three-dimensional (3-D) imaging design software (Straumann® Cares® Visual 6.0, Andover, Mass). The scanned information was transferred electronically to the production facility (Straumann, Andover, Mass). ZCAs were standardized to be .7 mm and 1 mm in thickness and in two different angulations (Straight angle and 15 degree angle).

Group II:

 Forty Implant replicas (n= 40) (10 for each subgroup) [Nobel Biocare, Nob

RpL RP 4.3×11mm, REF 29502 LDT436479, Yorba Linda, CA] were

placed in cubic self cure acrylic jigs [DENTSPLY Caulk, Caulk®

Orthodontic Resin] same as group I.

Zirconia Custom Abutments group II*:*

The implant replicas which were placed in cubic acrylic jigs were scanned in

order to design custom abutments digitally at the Prosthodontics Department

(Tufts University School of Dental Medicine) by a single operator using a

surface scanner (NobelProcera™ Scanner, Yorba Linda, Calif), Figure 8.

This scanner uses a laser beam to trace abutment position locator (RP 35551

Nobel biocare, Yorba Linda, CA) and render a digitized image of the

implant analog, It then designs the custom abutment digitally. The finish line

was set and adjusted as necessary using three-dimensional (3-D) imaging

software (NobelProcera® 3D GUI, Yorba Linda, Calif). The scanned

information transferred electronically to the production facility (Nobel-

Figure 8. A, B, C, and D scanning process for group II

-Biocare, Yorba Linda, CA). ZCAs were standardized to be .7 mm and 1mm

in thickness with two different angulations (straight angle, and 15° degree

angle), Figure 1.

Testing of fractural strength:

After receiving the ZCAs from the manufacturers, The ZCAs were engaged

to the implant replicas into the cubic acrylic jigs by using a manual torque

wrenchs prosthetic (L 84 Stainless steel, Straumann, Andover, Mass and

Nobel Biocare stainless steel L84, Yorba Linda, Calif) The torque wrench

was used to tighten abutments to the desired torque measurement based on manufacturer's recommendations (35Ncm for Group I and group II), Figure 9.[122, 123]
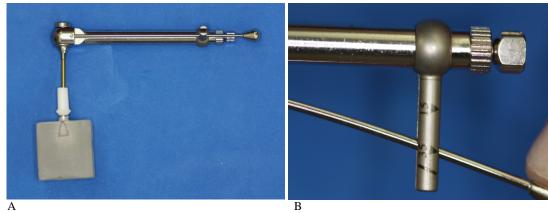

A   B
Figure 9. A & B torque wrench used to secure the samples

Figure10. Metallic Vice

Each jig was secured and mounted in metallic vice, Figure 10. The mounted jigs adjusted at 30 degrees relative to mechanical indenter for groups I-A-1, I-B-1, II-A-1, and II-B-1, Figure 11. However, the mounted jigs were adjusted at 15 degrees for groups I-A-2, I-B-2, II-A-2, and II-B-2, Figure 11.[124-129]

A   C

B   D
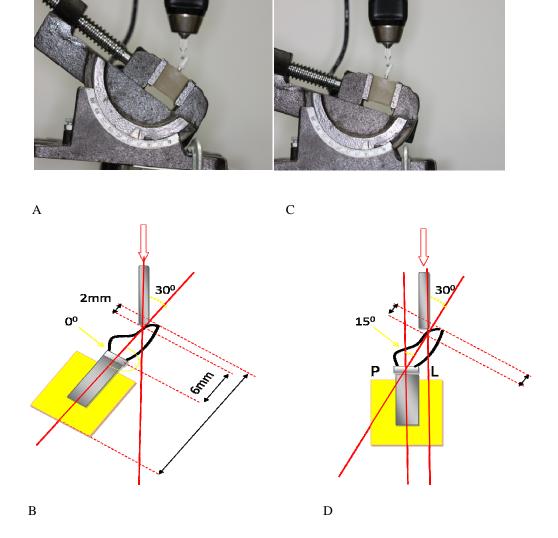
Figure 11. (A, B) Metallic vice adjusted to be 30°   Figure 12. (C, D) Metallic vice adjusted to be 15°

The indenter was covered by a resilient material (durasoft; scheu Dental GmbH, Iserlohn, Germany). The indenter contacts the entire mesio-destal occluding surface in a contact width of approximately 2-4 mm. The resilient material is a co-extrusion compound material consisting of a hard polycarbonate base and soft polyester urethane. It has been used to reduce localized contact stress intensities, and to distribute stress over the complete testing unit, including screws and abutments.[121] After that, the shear stress was measured by loading the samples until failure with a crosshead speed of .5 mm/min with the force transferred to the lingual surface of the ZCA 2 mm below the incisal edge by using Universal Testing Machine (Model 5566; Instron, Canton, Mass), Figure 13. The universal testing machine was controlled via a computer software system (Bluehill®2 Software, Mass), which also completed the stress-strain diagram and recorded the breaking load.

Figure 13*: Universal testing machine.*

Samples Size and Power Calculation:

For the sample size calculation, some studies related to the present study were reviewed and the closest study was chosen to measure the strength of the relationship between variables (effect size).[89, 146] With a sample size of 80 (10 per group) 80 percent power was achieved to detect a difference among the groups. Statistical software (R 2.11.1) was used to calculate the sample size required to achieve an alpha = .05 and a power of 80 percent.

Statistical Analysis:

Descriptive statistics have been reported for each group (mean, standard deviation, minimum values, and maximum values). A two-way analysis of variance was performed separately for groups I and group II to assess the statistical significance of each factor. Then, four independent samples t-tests were performed in order to compare group I and group II under each possible condition (thickness and angulations); even though we knew that the results might not be reliable due to the fact that the ZCAs were from

different manufactures. A Kolmogorov-Smirnov test was performed to check the normal distribution of residuals in group I and group II.

Results:

In group I, the mean fractural strength of groups A-1 through B-2 ranged from 162.49 ± 63.21 N to 231.14 ± 95.21 N (Table 2), while in group II, the mean fractural strength of groups A-1 through B-2 ranged from 431.83 ± 96.74 N to 745.93 ± 274.79N (Table 3).

Table 2. **Fractural strength (individual values, means, SD, minimum, and maximum values, in newton) in group I**				
Groups	**A-1**	**A-2**	**B-1**	**B-2**
Means	226.59	162.49	231.14	168.13
SDs	73.37	63.21	95.21	59.18
Min	141.09	104.38	111.26	108.78
Max	360.42	313.60	406.01	307.68

In group I, a Kolmogorov-Smirnov test was performed to check the normality distribution of the residuals across the groups. In this group, the p-value was p=.301, which means there was no evidence that the assumption

of normal distribution of the residuals is violated. Based on this, a two-way ANOVA was performed for this group.

In group I-A-1 and group I-B-1, one-piece straight abutments groups showed a higher fractural strength compared to group I-A-2 and group I-B-2 angulated one–piece ZCAs. The mean fractural strength was respectively from higher to lower 231.14 ± 95.21(B-1), 226.59 ± 73.38 (A-1), 168.13 ± 59.18 (B-2), 162.49 ± 63.21 (A-2). When the test was conducted for the first time, the interaction was not significant between the variables (p= .981), so the test was performed for second time without the interaction, and the one-piece straight ZCAs exhibited a significantly higher fractural strength than angulated ZCAs (p= .009). The box plots below represent the data that shows the difference between groups and the load that was applied to each group, Figure 14. No statistically significant difference was found between groups with different thicknesses, group I-A-1 & group I- A-2, and group I-B-1 & group I-B-2 where p=.827.
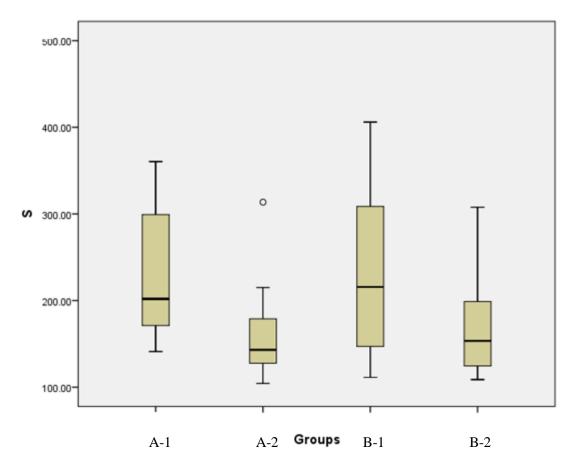
Figure 14 - Fractural strength (S) in Newton for group I (A-1, A-2, B-1 & B-2).

Table 3. **Fractural strength (individual values, means, SD, minimum, and maximum values, in newton) in group II**				
Groups	**A-1**	**A-2**	**B-1**	**B-2**
Mean	431.83	586.54	643.05	745.93
SD	96.74	187.54	193.36	274.79
Min	272.83	434.57	249.77	462.43
Max	598.27	1022.00	997.86	1233.49

In group II, a Kolmogorov-Smirnov test was used to test the normality distribution of residuals. There was no evidence that the assumption of normal distribution of the residuals is violated, (p=.509) which based on this, a two-way ANOVA was performed for this group. Two-piece angulated abutment group II-A-2 and group II-B-2 showed a higher fractural strength compared to group II-A-1 and group II-B-1 straight two–piece abutments. The mean fractural strengths were respectively from higher to lower 745.93 ± 274.79 (B-2), 643.05 ± 193.36 (B-1), 586.54 ± 187.54 (A-2), and 431.83 ± 96.74 (A-1). When the test was conducted for the first time, the interaction was not significant between the variables (p= .682), so the test was performed for the second time without the interaction, and it was found that the two-piece 1 mm thickness ZCAs exhibited significantly higher fractural

strength compared to .7 mm thickness ZCAs (p= .005). The box plots

represent the data that show the difference between the groups and the load

that was applied to each group, Figure 15. Also, there were statistically

significant differences found between groups with different angulations

(p=.045). Group II-B-2 and group II-A-2 showed a higher fractural strength

than group II-B-1 and group II-A-1. The maximum fractural strength was
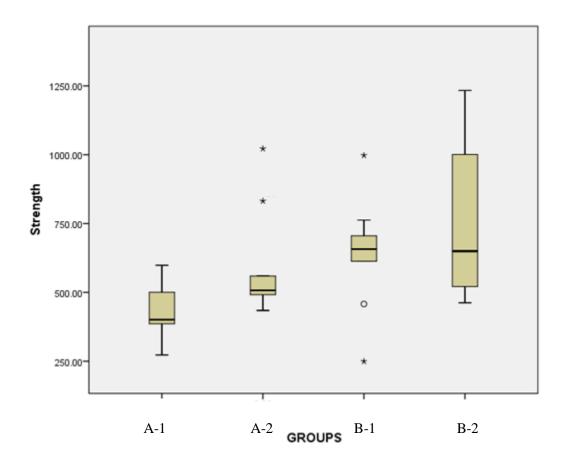
achieved in group II-B-2.

Figure 15 - Fractural strength (S) in Newton for group II (A-1, A-2, B-1 & B-2).

In order to compare the two main groups (Group I and group II) four different independent–samples t-tests were performed (Table 4) under different conditions (angulations and thickness).

Table 4. Summary of the tests which have been performed to compare group I (GI) & group II (GII)				
Test	GI vs GII	Thickness	Angulations	P-value (0.05)
1	GI-A-1 vs GII-A-1	.7 mm	0°	<0.001
2	GI-A-2 vs GII-A-2	.7 mm	15°	<0.001
3	GI-B-1 vs GII-B-1	.7 mm	0°	<0.001
4	GI-B-2 vs GII-B-2	.7 mm	15°	<0.001

Group I-A-1 vs group II-A-1: Levene's test was performed to check for equality of variances; equal variances were assumed (p=.513). Group II-A-1 showed a higher fractural strength (Mean=431.83 and SD=96.74) compared to group I-A-1 (Mean=226.59 and SD=73.37). There was a statistically significant difference between group I-A-1 and group II-A-1, with same thicknesses and angulations p<.001. The box plots containing whiskers and

outliers that represent the fractural strength (S) in newton and groups with different thicknesses and angulations showed in, Figure 16.

Group I-A-2 vs group II–A-2: Levene's test was performed to check for equality of variances; equal variances were not assumed (p=.036). Group II-A-2 showed a higher fractural strength (Mean=586.54 and SD=187.54) compared to group I-A-2 (Mean=162.49 and SD=63.21). There was a statistically significant difference between group I-A-2 and group II-A-2 with the same thicknesses and angulations (p<.001).

Group I-B-1 vs. group II-B-1: Levene's test was performed to check for equality of variances; equal variances were assumed (p=0.331). Group II-B-1 showed a higher fractural strength (Mean=643.05 and SD=193.36) compared to group I-B-1 (Mean=231.14 and SD=95.205). There was a statistically significant difference between group I-B-1 and group II-B-1 with the same thicknesses and angulations (p<.001).
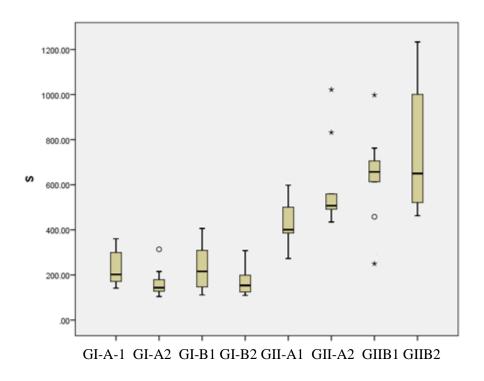
Group I-B-2 vs. group II-B-2: Levene's test was performed to check for equality of variances; equal variances were not assumed (p<.001). Group II-B-2 showed a higher fractural strength (Mean=745.95 and SD=274.79) compared to group I-B-2 (Mean=168.13 and SD=59.18). There was a statistically significant difference between group I-B-2 and group II-B-2 with the same thicknesses and angulations (p<.001).

Figure 16 - Box plots with whiskers and outliers of the load strength (S) in newtons, and the 8 different groups

Discussion:

Researchers highlighted the importance of tow-piece and one-piec ZCAs. [130-133] The majority of published investigations on all-ceramic implant abutments made from zirconia dioxide, examined simulated single incisor replacements. [134-137] These papers reported fractural strengths between 429 to 793 N under load angles that ranged from 30° to 60°.[135-137] Strong correlation exist between measured fractural strengths and the type of implant-abutment connection.[138] In the present study, the mean fractural strength for 0° and 15° in group I (one-piece ZCAs) ranged from 162.49 ± 63.21 N to 231.14 ± 95.21 N. And in group II (two-piece ZCAs) ranged from 431.83 ± 96.74 N to 745.95 ± 274.79 N. A simple bar graph was used to represent the differences between the means, Figure 17.
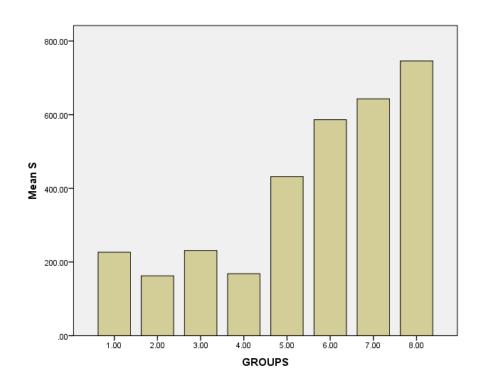
Figure 17. Simple bar graph representing the difference between the means.

Due to the various range of angles and difference in force application, in addition to different study designs, it is difficult to compare our values for fractural strength with results from other studies.[121-140]

The moment of force (Torque moment) played an important role in the present study; and had an important effect on the fractural strength of the ZCAs. The strength of the specimens might be affected by three quantities;[121] the force applied to the specimens, the length of the lever

arm connecting the axis to the point of force application, and the angle

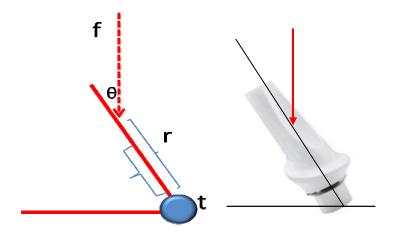between the force vector and the lever arm, Figure, 18.

Figure 18. $\boldsymbol{\tau} = \mathbf{r} \times \mathbf{F}$, $\tau = rF \sin \theta$

$\boldsymbol{\tau}$ = is the torque vector and τ is the magnitude of the torque,
$\mathbf{r}$ =is the displacement vector (a vector from the point from which torque is measured to the point where force is applied), and r is the length (or magnitude) of the lever arm vector,
$\mathbf{F}$= is the force vector, and F is the magnitude of the force,
θ =is the angle between the force vector and the lever arm vector.

For one-piece ZCAs the length of the lever arm victor was 12.6 mm, with

thickness of .4 mm and .6 mm in the area of implant-abutment connection,

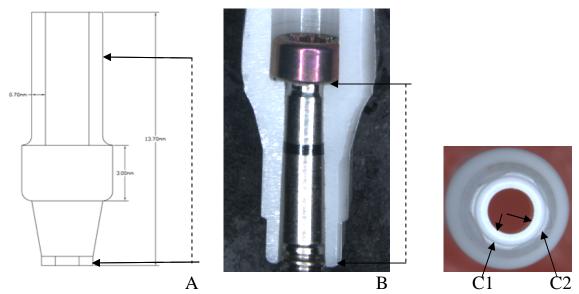Figure, 19. This thickness designed by the manufacture and is standard and

unchangeable.

Figure 19: **A**; the length of the lever arm vector =12.6 mm. **B**; The length of the implant-abutment connection = 5.5 mm. **C**; the thickness of implant abutment connection which in C1= .6 mm, and C2= .4 mm

The thinnest part in one-piece ZCAs is in the implant-abutment connection

area (.4 mm and .6 mm). However, this area is the most critical area in the

abutments because of the force concentration which might be one of the

reasons of the failures in this area. Based on this finding one can interpret

that the thickness of ZCAs in this design had no influence on the fractural

strength since the thickness of implant-abutment connection is standard.

However, the anglulation had a negative influence on the fractural strength

because when the force was applied on angulated abutments the angle (θ)

between the force vector (f) and the lever arm (r) was increased, which

decreased the force required to break the specimens which could be one of

the reasons for failure in this design. Whereas, in two-piece ZCAs the length

of lever arm victor is 6 mm (which much shorter than one-piece ZCAs). In

addition, the thickness of the ZCAs had a statistical significant and positive

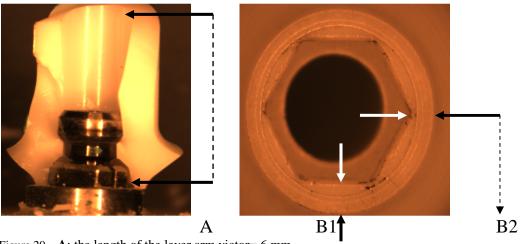influence on the strength of the two-piece ZCAs, Figure 20.

A   B1   B2

Figure 20. **A**; the length of the lever arm victor= 6 mm
    B; the thickness of implant-abutment connection which in B1= .7 mm and B2= .7 mm

These findings gave an explanation that the angulation is significant in two-

piece ZCAs. The angle (θ) [between the force vector (f) and the lever arm

(r)], did not change when force was applied to angulated abutments, which

can be one of the reasons for high fractural strength achieved by this design.

In this study, static loading was applied slowly with a crosshead speed of .5

mm/min. This corresponds to the load in a para-functional situation rather

than chewing. Ferrario et al, measured single-tooth bite forces in healthy young adults and they reported results of 150 N and 140 N for the central and lateral incisors respectively in men.[139] Higher bite forces are to be expected in subjects with functional disorders, such as bruxism.[140] In the present study the means of fractural strengths a cross the groups exceeded the average biting force (140 N to 150 N).[140] However, we are uncertain in predicting the performance of one-piece ZCAs in individuals with functional disorders. We emphasize the need to pay special attention to the occlusal relationship of the lower and upper jaw whenever possible, and we recommend keeping such abutments free from dynamic occlusion. The two-piece angulated ZCAs exhibited higher mean fractural strength compared to straight abutments in two-piece ZCAs, which were statistically significant. This data leads us to reject our null hypothesis. In one-piece ZCAs, stress concentration was higher in the area of the apical hexagon, which is the thinnest portion of the abutment. In this study also artificial dynamic thermal ageing was not applied to the specimens since in previous studies artificial

aging failed to exert a statistically significant influence on the fractural strength of either straight or angulated abutments. [141-146] Static loading was performed at an angle of 30° to the long axis of the abutments in order to simulate a worst-case scenario. In the present study, the most typical fracture pattern (95 percent of the samples in this group) in group I (one-piece ZCAs) was an oblique fracture line below the implant shoulder, Figure 21.
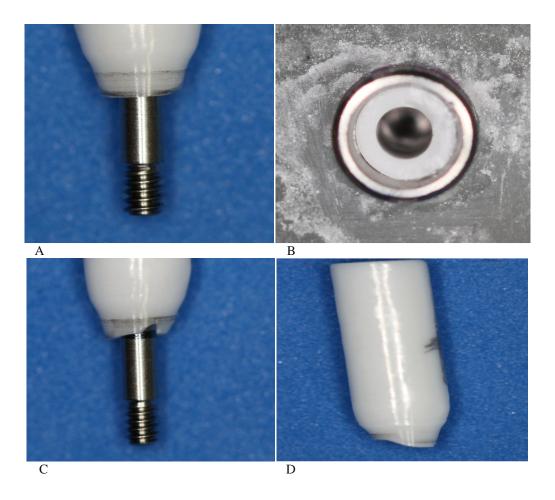
A

B

C

D

Figure 21: A, B, C, and D Fracture pattern appears as an oblique line below the abutment's shoulder.

These findings are in agreement with Adatia et al, indicates that certain grinding procedures above the level of the implant shoulder for customize the abutments have no impact on fractural resistance. [136] However, this claim may be only valid for one-piece ZCAs of implant-abutment connection tested in this study.

Two-piece ZCAs exhibited significantly higher fractural strength compared to one-piece ZCA internally connected abutments. Therefore, it can be concluded that the type of connection to the implant influences the stability of ceramic abutments regardless of the thickness of the ZCAs in group I, while the thickness had a positive influence on the stability of ZCAs in group II. However, there was a significant effect of the angulations on the strength of the ZCAs in group I and group II.

Metallic internal connections have also been associated with a more favorable load distribution in the connection area.[147] A finite element

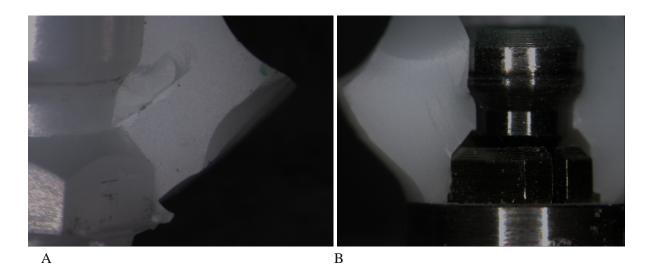A   B

Figure 22. Load distribution in two-piece ZCAs. A; crack propagate through ZCA in screw-metallic insert area. B; multiple cracks as result of force-distribution.

analysis demonstrated high tensile stress in the abutment screw threads upon lateral loading in the implant-abutment connection area, whereas in the metallic internal connection, lateral loading was taken up by the metallic connection, thus protecting the thread portion of the abutment from load transfere.[148]   The present study is in agreement with these findings. The results of this study demonstrated a superior distribution of the load with an internal implant- abutment connection for tow-piece ZCAs.

Additionally, the metallic insertion seems to be advantageous to transfer the forces comparing to one-piece ZCAs. The one-piece angulated ZCAs

exhibited the lowest fractural strength in the present study. The two-piece

ZCAs exhibited a fractural strength mean ranged between 431.83 ± 96.74 N

to 745.95 ± 274.79 N under static loading which was statistically significant.

In comparable investigation, the fractural strength of the same type of

abutments was 738 ± 245 N under static force application.[103] Generally, it is

difficult to compare the data between studies that related to the fractural

stability of ZCAs, because of different study designs and the direction of

load application with different lever arms. Furthermore, variations in the

angle of the applied load, static or dynamic testing methods, the size and

shape of the abutments, could all have an impact on the results.

The fractural strength of the abutments with metallic inserts was within the

range of the values reported in other studies. [108, 109] In these studies, the

stability of zirconia, alumina, and titanium abutments was analyzed after

connection to the same type of implants.[108, 109]
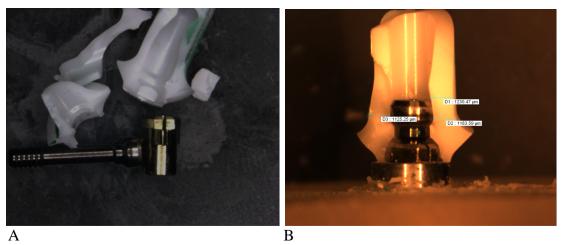
 A   B

Figure 23. A & B: plastic deformation of the metallic components

In two-piece ZCAs groups, plastic deformation of the metallic insertions and

abutment screws of the samples were observed, Figure 23.

That could be one of the reasons for the higher standard deviations in these

groups. In group II, the failure pattern encompassed the fracture of the

zirconia abutments, and/or deformation of the metal parts. A variation in the

fracture pattern was also observed in other studies reporting on alumina,

zirconia, and titanium abutments with internal connection.[108,109] In only one

of those studies, however, an implant neck distortion was found in a

specimen bearing a titanium abutment.[108] The abutment screw was identified

as the weakest component in previous study examining titanium

abutments.[109] In another investigation testing alumina and zirconia

abutments, fracture of the abutment and/or fracture of both the abutment and

crown were the main reasons for failure.[108] In contrast to these results, in the

present study, only ceramic components failed by fracture in both groups

with ceramic or metal insertion, but the fracture was significantly higher in

the groups with ceramic insertion . One reason for the difference found in

these investigations might be associated to the difference in force

application.

Whereas dynamic loads were used in the two previous studies, static loads

were applied in the present study, allowing higher loads before failure.

Nevertheless, naturally occurring forces in patients remain far below the

forces recorded in these *in vitro* studies.[104] In clinical situations, therefore, a

plastic deformation of the metallic components is unlikely to occur.

Limitations:

-The present study have been conducted as *in vitro* study and there was no control group

- The specimens have been tested without artificial crowns.

- There was no artificial cyclic loading.

Conclusions:

Within the limitations of this *in vitro* study, the following conclusions can be drawn:

- ZCAs with secondary metallic component exhibited the highest fractural strength, especially with angulated abutments.

- One-piece ZCAs were the lowest of those investigations, especially the angulated abutments.

- The thickness of two-piece ZCAs had a positive influence on the fractural resistance but not in one-piece ZCAs.

List of tables

List of figures

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