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**Comparison between the fracture strength and
failure mode of lithium disilicate, zirconia and
titanium implant abutments**

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ABBREVIATIONS

CAD: Computer Aided Design

CAM: Computer Aided Manufacturing

GPa: Gigapascal

LTD: Low Temperature Degradation

mm: Millimeter

MPa: Megapascal

N: Newton

Y-TZP: Yttrium-Stabilized Tetragonal Zirconia Polycrystal

1. INTRODUCTION

1.1. Dental implant restorations

Implant-supported single restorations have been a valid treatment alternative to conventional prostheses for the replacement of missing teeth. High success rates of 93.7% of implant-supported crowns after an observation period of at least 5 years are documented ^[7]. In addition, survival rates are comparable to conventional fixed dental prostheses retained by crowns on natural teeth ^[62, 103].

The goal to be achieved in implant dentistry is not just to place an implant, but to restore functions and esthetics of a missing tooth. Thus, the success of the implant restorations does not depend only on osseointegration and function, but also on achieving natural and harmonious appearance of the replaced missing teeth, which depends on the materials used for both, the implant abutment and the crown.

Titanium abutments restored with porcelain fused to metal crowns have been known to be the standard treatment option in implant dentistry with high survival rates ^[80, 83, 133], due to the high mechanical properties and biocompatibility ^[2, 125]. However when using titanium, the esthetic result of the final restoration can be compromised through a gray color which may be transmitted through the periimplant tissues giving an unnatural bluish appearance, especially in cases with a thin tissue biotype or inadequate depth of the emergence profile ^[58, 79, 105]. In addition, when titanium abutments are restored with all-ceramic crowns, the underlying metal abutment receives a certain percentage of incident light which can change the final color of the restoration ^[50, 120, 129].

To achieve optimal esthetics, new generations of all-ceramic abutments restored with all-ceramic crowns have been developed to prevent the unnatural metallic color of titanium. One limitation of ceramics is the high brittleness and their potential to crack ^[9]. This is always a concern when using all-ceramic abutments and whether they could withstand functional forces in the oral cavity.

1.2. All-ceramic abutments

Ceramics can be divided according to the chemical composition into three categories: 1) Non-oxide ceramics such as nitrides and carbides; 2) non-silicate or high strength oxide ceramics such as alumina or zirconia; 3) silicate ceramics, which could be further differentiated into feldspathic porcelains and glass ceramics (such as lithium disilicate ceramic) [23, 95].

Initially in the nineties of the last century, all-ceramic abutments were fabricated out of densely sintered high-purity alumina (Al_2O_3) ceramic. Since 2000 zirconia ceramic has been also used as abutment material instead of alumina. Recently, lithium disilicate glass ceramic has been introduced as an abutment material as it can provide more natural esthetics when compared to oxide ceramics [18, 56, 77, 82, 115].

All-ceramic abutments offer several clinical advantages over metal abutments. Ceramic abutments provide better esthetics and significantly contribute to a lower discoloration of the mucosa than metal abutments [62]. In addition, bacterial adhesion on ceramics such as zirconia was found to be less than on titanium [113]. Finally, the soft tissue integration of zirconia ceramic is similar to that of titanium [73].

One shortcoming of ceramics is their brittleness, therefore, they show less resistant to tensile forces than metals. Micro-structural defects within the material may cause cracks in combination with tensile forces [20]. An increase in the fracture toughness of a ceramic slows down crack propagation and consequently has a major influence on the material's long-term clinical stability [34].

1.2.1. Zirconia abutments

Zirconia is a polycrystalline ceramic without any glass component. It can be found in three crystallographic phases: (1) the monoclinic phase at room temperature that is stable up to $1,170^\circ\text{C}$, (2) the tetragonal phase that starts at $1,170^\circ\text{C}$ and is stable up to $2,370^\circ\text{C}$ and (3) the cubic phase that occurs above $2,370^\circ\text{C}$. When the zirconia is not stabilized, a transformation of tetragonal to monoclinic phase occurs while cooling down to the temperature of $1,170^\circ\text{C}$. By the addition of stabilizing oxides to pure zirconia, such as calcium oxide (CaO), magnesium oxide (MgO), cerium oxide (CeO_2) or yttrium oxide (Y_2O_3), material's phase transformations can be inhibited and zirconia can be stabilized in its tetragonal phase at room temperature, termed as stabilized zirconia [32]. Tensile stresses at a crack tip will cause the tetragonal phase to transform into the monoclinic phase with an associated 3-5% localized expansion [102]. This

volume increase creates compressive stresses at the crack tip that counteract the external tensile stresses and retards crack propagation. This phenomenon is known as transformation toughening^[17]. As a result of transformation toughening, zirconia ceramics exhibit high flexural strength (900 to 1,200 MPa), compression strength (2,000 MPa) and fracture toughness (ranging from 6 to 8 MPa.m^{0.5})^[34, 39, 59, 84, 89, 130].

Due to the well-documented high fracture resistance, good esthetics and superior biocompatibility, zirconia ceramic has recently attracted significant interest which led to its use as implant abutment^[10, 49, 69, 113]. Zirconia abutments manufactured using computer-aided design/computer-aided manufacturing (CAD/CAM) technology is one of the most popular treatment options in implant dentistry especially in the esthetic zone^[13, 107].

Zembic et al. showed excellent survival rates for single-implant all-ceramic crowns supported by zirconia abutments and estimated 5-year survival and failure rates comparable to metal abutments^[131]. Loosening of the abutment screw was one of the few technical complications occurring at zirconia abutments^[49]. This finding is similar to the observations made with metal abutments. Moreover, a survival rate of 96.3% for zirconia abutments with all-ceramic crowns in anterior and premolar regions as reported after eleven years^[132]. Kim et al. reported a 5-year success rate of 95% for 611 alumina-toughened zirconia abutments used to support 328 implant-supported fixed restorations (60 in anterior region and 268 in posterior region)^[71].

Hosseini et al. studied the biological, technical and esthetic outcomes of treatments with implant-supported, single-tooth restorations consisting of metal or zirconia abutments and metal-ceramic or all-ceramic crowns, and reported high implant survival rates and few biological complications that occurred during 3 years. However, more marginal bone loss was registered with gold alloy compared to zirconia abutments^[60].

Good esthetic results can be achieved through manufacturing of abutments entirely made of zirconia, yet, weak and fracture-prone points might develop at the connection especially with internal connection types^[3, 33]. Moreover, an imprecise fit between the abutment and the implant can result in case of all-ceramic abutments, as they cannot be machined to the same degree of precision as metal abutments. An imprecise fit between abutment and implant can lead to screw loosening as well as successive microbial infections which may lead to bone loss^[22].

Implementing a titanium insert into zirconia abutments can overcome these problems and improve the fracture resistance of the abutment [41]. Chun et al. recently tested both types of zirconia abutments and found that the internal friction connection that used a titanium insert with a zirconia abutment was more resistant to loading than the internally connected pure zirconia abutment [33]. The use of a titanium insert provides more support to brittle ceramics and provides a more precise fit with the implant. This avoids the weakest point of the zirconia abutment at the implant-abutment contact area, and the undesirable color of the metal can then be masked with the zirconia suprastructure. Such an assembly makes use of both advantages of metal and zirconia abutments. Truninger et al. found that zirconia abutments with internal implant–abutment connection used with a secondary metallic component exhibited the highest bending moments compared to pure zirconia abutments (with no titanium insert) and also compared to zirconia abutments with an external implant–abutment connection [122]. This shows that the type of implant–abutment connection and the use of secondary metallic component can influence the mechanical properties of zirconia abutments. In a recent study, when five different zirconia abutments were tested for fracture strength, zirconia abutments with titanium inserts again demonstrated a greater fracture resistance than the pure zirconia abutments [127].

The whitish color of the zirconia abutment offers favorable esthetics compared to the grayish color of titanium in clinical situation of thin peri-implant mucosa or all-ceramic crowns [63, 93, 111, 123]. Nonetheless, the zirconia core is less translucent than other dental all-ceramic materials such as glass-ceramics [40, 56]. Lithium disilicate containing glass-ceramics have proven to be successful esthetic options compared to zirconia which has poorer translucency and that often is too white for an optimal esthetic appearance [1, 18, 56].

1.2.2. Lithium disilicate abutments

The strongest and toughest glass ceramic available is lithium disilicate. It has a moderate flexural strength (360–440 MPa) [6] and fracture toughness (2.5–3 MPa.m^{0.5}) [52], excellent translucency and shade matching options [56, 57]. Higher translucency is observed for lithium disilicate ceramic than for zirconia ceramics [18, 78]. Lithium disilicate is available in the market in a pressable (IPS emax Press) and machinable form (IPS emax CAD). Lithium disilicate glass-ceramic has shown promising results in terms of structural integrity when used in anterior or posterior area as inlays [43, 104], crowns [48, 90], partial coverage restorations [53], and fixed dental prostheses [28].

In esthetic demanding cases the tooth-colored lithium disilicate offers better and more natural esthetics than the less translucent zirconia abutments [18, 56, 57]. Improved material characteristics, and complying with increased clinician and patient demands for highly esthetic results, contributes significantly to the possibility of usage of lithium disilicate abutments as alternative to zirconia in the esthetic zone.

There are two possibilities of using lithium disilicate abutments; as a hybrid abutment bonded on titanium insert and on top of it an all-ceramic crown, or as hybrid-abutment-crown where the abutment and crown are manufactured as one piece that is bonded to a titanium insert and screwed to the implant; the screw hole is subsequently closed with composite material [77, 82, 115].

As the dilemma of cement-retained or screw-retained is not solved in the literature, the choice of retention in fixed implant prosthodontics seems to be based mainly on the clinician's preference [27, 121]. The main advantages of screw-retained restorations are retrievability [30, 129] and excellent marginal integrity [64]. Nevertheless, screw-retained restorations are associated with more sophisticated clinical and laboratory procedures, which increase the total cost of implant treatment [29, 92]. Cement-retained restorations can be used to eliminate such disadvantages as well as to broaden the material spectrum to include additional restorative materials, i.e. all-ceramic materials [121]. However, several disadvantages have been mentioned in the literature for such restorations, such as the difficult retrievability [55, 66], in addition to the demanding removal of excess cement. Peri-implant tissue inflammation can be caused by residual cement in the soft tissues [4, 100]. Another disadvantage of cement-retained restorations is the reduced stability in situations where interocclusal space is limited, as the abutment lacks the important factors of height and surface area for cement retention [27, 42].

The use of hybrid-abutment-crowns can eliminate these problems. It allows ease of access to the screw through the composite, moreover, it eliminates the crown margins and the need for cementation. Nonetheless, the need for an optimal surgical implant positioning is required, as an incorrect implant axis would lead to a hole in the labial surface of the incisors created by the screw cavity, which would be unacceptable due to the high esthetic requirements in this area.

To be considered a true alternative, the performance of lithium disilicate abutments must be comparable to the widely used titanium and zirconia abutments. Clinical and laboratory studies should be performed to prove materials' applicability and performance. Laboratory tests can be done in a short period of time and have the advantages of reproducibility and the

possibility of standardizing test parameters ^[67]. Well-designed laboratory tests can be an effective indicator of the reliability of the materials and its likely clinical success, before widespread clinical use is recommended.

Few articles and case reports are available regarding the use of lithium disilicate ceramic as a material for implant abutments ^[77, 82, 115]. However, data regarding the fracture strength of lithium disilicate as single unit implant abutments are missing in the literature. Additionally, limited laboratory and clinical studies compare the mechanical properties of pure zirconia abutments with the hybrid zirconia abutments supported by titanium inserts using internal conical connections, and showed higher fracture strength for the hybrid zirconia abutments ^[33, 122].

2. AIM OF THE STUDY

The purpose of this study was to:

1. Evaluate the fracture resistance of implant-supported all-ceramic crowns on zirconia or lithium disilicate ceramic abutments and to compare them with all-ceramic crowns on conventional titanium abutments.
2. Identify the mode of failure of the aforementioned restorations.
3. Evaluate the effect of titanium inserts on the fracture strength of crowns on zirconia abutments.
4. Test and compare hybrid-abutment-crowns as a new treatment modality.

The null hypothesis to be tested was that there is no difference in load to fracture between the three different abutment materials, as well as different forms of using zirconia or lithium disilicate abutments.

3. MATERIALS AND METHODS

3.1. Materials

The materials used in the study and their batch numbers are shown in Table 1. All restorations used have been adhesively cemented, and all materials used for bonding and its composition are described in Table 2 and shown in Figure 1.

Table 1: List of materials used in the study

Material	Manufacturer	Generic Name	LOT No.
FairTwo Implants	FairImplant, Bönningstedt, Germany	Grade 4 titanium implants	135395
FairTwo Abutment scanbase S	FairImplant	Grade 4 titanium inserts	399275
FairTwo Abutment screw	FairImplant	Titanium screw	961209
CopraTi-4 Titanium	Whitepeaks Dental, Essen, Germany	Grade 4 titanium blanks	T4001
Wieland Zenostar	Wieland Dental, Pforzheim, Germany	Zirconia blanks	S23252
IPS e.max CAD	Ivoclar Vivadent	Lithium disilicate glass ceramic	T44702
Multilink Automix	Ivoclar Vivadent	Dual-curing luting composite resin	T29844
Multilink Hybrid Abutment	Ivoclar Vivadent	Self-curing luting composite resin	T35016
IPS Ceramic Etching Gel	Ivoclar Vivadent	Etching gel	T40413
Monobond Plus	Ivoclar Vivadent	Universal primer	S14727
Liquid strip	Ivoclar Vivadent	Glycerin gel	T29465
Tetric EvoCeram	Ivoclar Vivadent	Composite resin	T09635

Technovit 4000 -Powder -Technovit Syrup I -Technovit Syrup II	Heraeus Kulzer, Wehrheim, Germany	Self-curing polyester resin	010294 011065 012033
Chlorohexamed	GlaxoSmithKline, Bühl, Germany	Antiseptic gel	4233060
Steatite Ceramic Ball	Höchst Ceram Tec, Wunsiedel, Germany	Steatite Ceramic	-
Zwick Z010	Zwick, Ulm, Germany	Universal testing machine	-
Chewing Simulator CS-4	Chewing Simulator CS-4, SD- Mechatronik, Westerham, Germany	Chewing Simulator	-



Figure 1: Materials used during the bonding procedures

Table 2: Materials used for bonding and their composition

Material	Composition
Multilink Hybrid Abutment	<p><u>Standard composition: (in wt%)</u></p> <p>Base:</p> <ul style="list-style-type: none"> ● Ytterbium trifluoride 20-<25% ● Ethoxylated bisphenol A dimethacrylate 10-<25% ● Bis-GMA 10-<20% ● 2-Hydroxyethyl methacrylate 2.5-<10% ● Titanium oxide 1-<2.5% ● 2-Dimethylaminoethyl methacrylate 0.1-<1% <p>Catalyst:</p> <ul style="list-style-type: none"> ● Ytterbium trifluoride 10-<25% ● Ethoxylated bisphenol A dimethacrylate 10-<25% ● Urethane dimethacrylate 3-<10% ● 2-Hydroxyethyl methacrylate 3-<10% ● Dibenzoyl peroxide 1-<2.5%
Multilink Automix	<p><u>Standard composition: (in wt%)</u></p> <ul style="list-style-type: none"> ● Dimethacrylate and HEMA (Base: 30.5 – Catalyst: 30.2) ● Barium glass filler and Silica filler (Base: 45.5 – Catalyst: 45.5) ● Ytterbiumtrifluoride (Base: 23.0 – Catalyst: 23.0) ● Catalysts and Stabilizers (Base: 1.0 – Catalyst: 1.3) ● Pigments (Base: < 0.01 Catalyst: -)
IPS Ceramic Etching Gel	Hydrofluoric acid: 2.5-<7%
Monobond Plus	Alcohol solution of silane methacrylate, phosphoric acid methacrylate and sulphide methacrylate
Liquid strip	Glycerin gel
Tetric EvoCeram	<p><u>The monomer matrix:</u></p> <ul style="list-style-type: none"> ● Dimethacrylates (17–18% weight) <p><u>The fillers:</u></p> <ul style="list-style-type: none"> ● Barium glass ● Ytterbium trifluoride ● Mixed oxide and prepolymer (82–83% weight) ● Additional contents: additives, catalysts, stabilizers and pigments

3.1.1. FairTwo Implants (FairImplant, Germany)

FairTwo implant (FairImplant) is a two-piece implant system that has an internal conical connection with a platform-switch feature. The implants are manufactured from grade 4 titanium alloy. According to the manufacturer, the implant can induce rapid bone formation and improved osseointegration due to the fact that the surface is covered with calcium phosphate layer (CaP) which has a hydrophilic effect and can absorb the blood once placed in the drilled bone site. The CaP layer is then completely resorbed, which leads to the increase of the bone-to-implant contact ratio and a strong bone-to-implant interface. The implant abutment is fixed into the implant with a conical connection, which is below the marginal bone level and therefore transfers functional loads deep down in the bone. The implants are commercially available in four diameters (3.5 mm, 4.2 mm, 5.0 mm and 6.0 mm) and five lengths (7.5 mm, 9.5 mm, 11.5 mm, 14.5 mm and 17.5 mm). The implants used in this study had a diameter of 4.2 mm and length of 11.5 mm.

3.1.2. CopraTi-4 Titanium (Whitepeaks Dental, Germany)

CopraTi-4 is a pure titanium grade 4 blank with the same positive properties like grade 2 titanium. It is also indicated for implants and construction elements as abutments and attachments. CopraTi-4 features high mechanical strength, therefore it is indicated for larger restorations. Also it is easy to mill as the chip removal properties are far better than grade 2 titanium. Any type of veneering porcelain for titanium can be used. CopraTi-4 consists of >99% Titanium (mass %). It has a density of 4.51 g/cm³, Vickers hardness 150 HV 5/30 and tensile strength of 345 MPa. Titanium abutments used in this study were milled from CopraTi-4 blanks.

3.1.3. Wieland Zenostar (Wieland Dental, Germany)

Zenostar is composed mainly of zirconia ($ZrO_2 + HfO_2 + Y_2O_3$) > 99.0 %, this includes yttrium oxide (Y_2O_3) > 4.5-6.0%, hafnium oxide (HfO_2) ≤ 5.0%, aluminum oxide (Al_2O_3) < 0.5% and other oxides < 0.5%. Zenostar is available in two translucency levels and in a variety of shades. The shade designations and the shading of Zenostar T and Zenostar MO are matched to the IPS e.max shade system. Consequently, Zenostar is the only zirconium oxide system that is fully compatible with IPS emax. This makes Zenostar abutment restored with IPS emax crown an advantage. The indication spectrum of the material ranges from crowns to multi-unit fixed dental prostheses (FDPs).

3.1.4. IPS e.max CAD (Ivoclar Vivadent, Liechtenstein)

IPS e.max CAD (Ivoclar Vivadent) is a lithium disilicate glass-ceramic block for the CAD/CAM technology. The blocks are available in three levels of translucency (HT, LT, MO) and two sizes (I12, C14). The material main component is SiO₂, additional contents are Li₂O, K₂O, MgO, Al₂O₃, P₂O₅ and other oxides. IPS e.max CAD crowns are recommended as single tooth restorations for all intraoral regions. The material is milled in crystalline intermediate stage then crystallization is done. The crystallization process at 840-850°C results in a transformation of the microstructure, during which lithium disilicate crystals grow in a controlled manner. The densification of 0.2% is accounted for in the CAD software and taken into account upon milling.

3.1.5. Multilink Automix (Ivoclar Vivadent, Liechtenstein)

Multilink Automix (Ivoclar Vivadent) is dual-curing luting composite resin (self-curing luting composite with light-curing option) that is commercially available in three shades (yellow, transparent, opaque) and can be used for the permanent adhesive cementation of different metal and ceramic (i.e. zirconia, lithium disilicate) indirect restorations such as inlays, onlays, crowns, FDPs and endodontic posts. It is composed mainly of hydrolytically stable phosphoric acids (acidic monomers). The monomer matrix is composed of 22 to 26% dimethacrylate, 6-7% HEMA and 1% benzoyl peroxide. The inorganic fillers (40% in volume) are barium glass, ytterbium trifluoride and spheroid mixed oxides with a particle size of 0.25-3.0 microns (mean particle size 0.9 microns).

3.1.6. Multilink Hybrid Abutment (Ivoclar Vivadent, Liechtenstein)

Multilink Hybrid Abutment (Ivoclar Vivadent) is a self-curing luting composite resin for the permanent cementation of ceramic structures made of lithium disilicate glass-ceramic or zirconium oxide on titanium/titanium alloy or zirconium oxide bases (e.g. abutment or adhesive basis) in the fabrication of hybrid abutments or hybrid abutment crowns. The material in the form of automix syringes and commercially available in two shades (HO 0 and MO 0). Multilink Hybrid Abutment is indicated only for laboratory use.

3.2. Methods

3.2.1. Study outline

Eighty single implant-supported restorations were assembled using eighty titanium implants with a diameter of 4.2 mm and length of 11.5 mm, having internal conical connection and platform-switch (FairTwo, FairImplant, Bönningstedt, Germany). Eighty ceramic crowns made from lithium disilicate glass ceramic (IPS emax CAD, Ivoclar Vivadent, Schaan, Liechtenstein) are produced to replace a maxillary right central incisor of 11 mm length and 8.5 mm width. For the purpose of this study, the specimens were standardized except for the abutment material, which differed between the test groups.

The implants were randomly divided, according to the abutment material and type, into five groups of sixteen implants each, then each group was divided into two subgroups (n=8) according to the type of load applied.

Titanium abutments (CoproTi-4, Whitepeaks, Essen, Germany) were used for the control group, **Ti**, whereas zirconia abutments (Wieland Zenostar, Wieland Dental, Pforzheim, Germany) were used with no titanium inserts and with titanium inserts for test groups **Zr** and **ZrT**, respectively. Lithium disilicate (IPS e.max CAD, Ivoclar Vivadent) were used as abutment in test group **LaT** and as hybrid-abutment-crown in test group **LcT**. For simplifying the groups and assembly used for each one, Table 3 summarizes different abutment-crown combination with their group codes. Moreover, Figure 2 shows the different parts used for the five groups.

Table 3: Overview of the abutment-crown combination in the five test groups

<u>Group</u>	<u>Subgroup</u>	<u>Abutment</u>	<u>Crown</u>	<u>No. of specimens</u>	<u>Fatigue load</u>
Ti	Ti1	Titanium ¹	Lithium disilicate ³	8	Yes
	Ti2	Titanium ¹	Lithium disilicate ³	8	No
Zr	Zr1	Zirconia ² (No metal insert)	Lithium disilicate ³	8	Yes
	Zr2	Zirconia ² (No metal insert)	Lithium disilicate ³	8	No
ZrT	ZrT1	Zirconia ² (Metal insert)	Lithium disilicate ³	8	Yes
	ZrT2	Zirconia ² (Metal insert)	Lithium disilicate ³	8	No
LaT	LaT1	Lithium disilicate ³ (Metal insert)	Lithium disilicate ³	8	Yes
	LaT2	Lithium disilicate ³ (Metal insert)	Lithium disilicate ³	8	No
LcT	LcT1	Lithium disilicate ³ hybrid-abutment-crown (Metal insert)		8	Yes
	LcT2	Lithium disilicate ³ hybrid-abutment-crown (Metal insert)		8	No

- (1) CopraTi-4 (Whitepeaks, Essen, Germany)
- (2) Wieland Zenostar (Wieland Dental, Pforzheim, Germany)
- (3) IPS e.max CAD (Ivoclar Vivadent, Schaan, Liechtenstein)

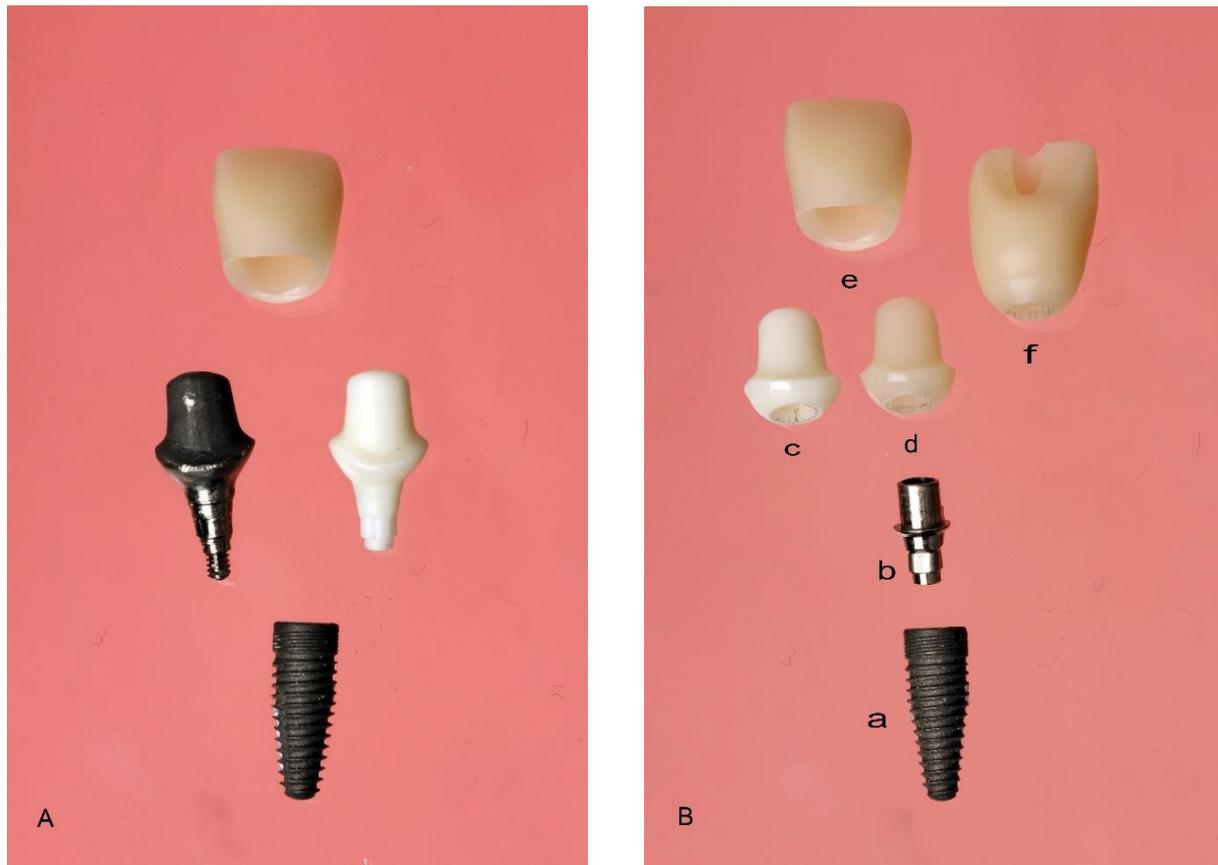


Figure 2: Restoration components of:

A: Groups **Ti** and **Zr**. Implants and crowns were identical between both groups, the only difference was the abutment; titanium on left, zirconia on right.

B: Restoration components of groups **ZrT**, **LaT** and **LcT**.

Group **ZrT**: Implant (a), titanium insert (b) bonded to zirconia suprastructure (c), and crown (e) was cemented on abutment.

Group **LaT**: Implant (a), titanium insert (b) bonded to lithium disilicate suprastructure (d), and crown (e) was cemented on abutment.

Group **LcT**: Implant (a), titanium insert (b) bonded to lithium disilicate hybrid-abutment-crown (f).

3.2.2. Abutment and crowns manufacturing

For the purpose of the study, standardization of the dimensions of the restorations was required. At first a crown of specific dimensions (11.0 mm in height and 8.5 mm in mesio-distal width) was planned. The minimum layer thickness was respected for each material used as abutment (zirconia and lithium disilicate) as well as the crown material. When the original titanium inserts were used, it was found to be difficult to achieve the required layer thickness if the crown was to be prepared. Therefore, the titanium inserts and the abutments were designed according to the needs of this study (Fig. 3).

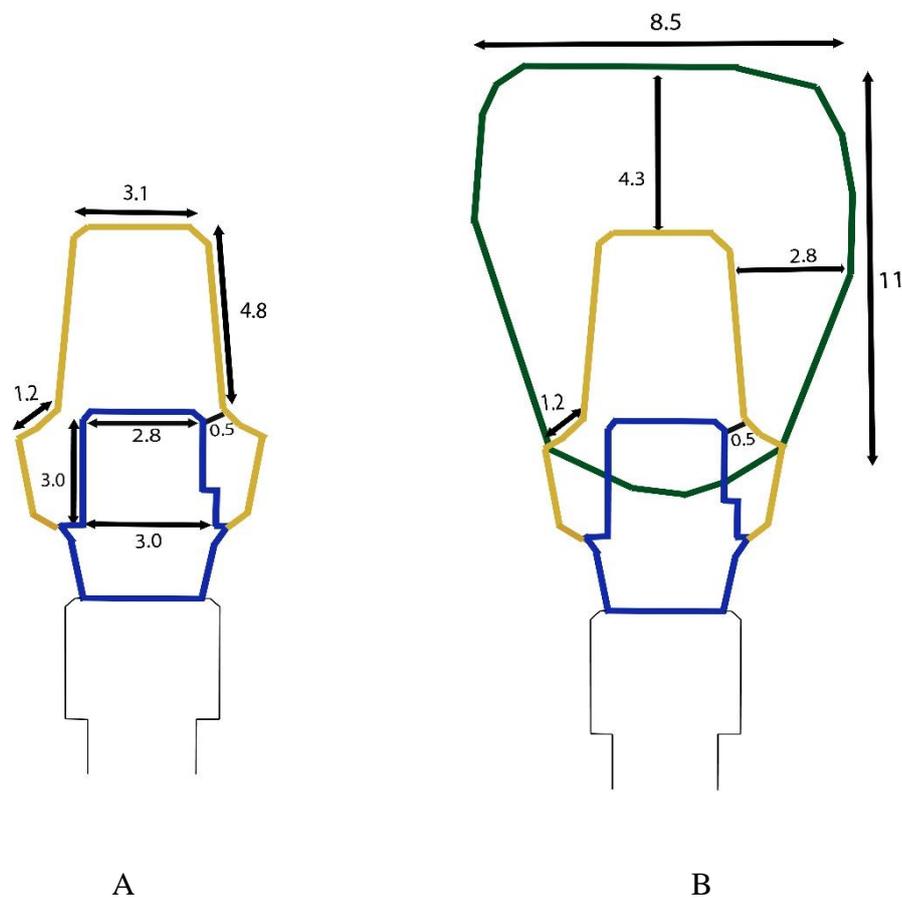


Figure 3: The design made to manufacture the abutments and crowns using a shortened titanium insert (3 mm). Titanium insert (blue), abutment/suprastructure (yellow), crown (green). All dimensions are given in mm.

A: Dimensions of the abutment and titanium basis.

B: Dimensions of the crown when seated on the abutment.

According to the manufacturer, minimum thickness of 0.5 mm was required for the IPS emax CAD abutments, and 1.0 mm (circumferentially) and 1.5 mm (incisally) for the crowns. Therefore, a silicone index was fabricated and cut to measure the available distance using rubber strips of known thickness as shown in Figure 4.

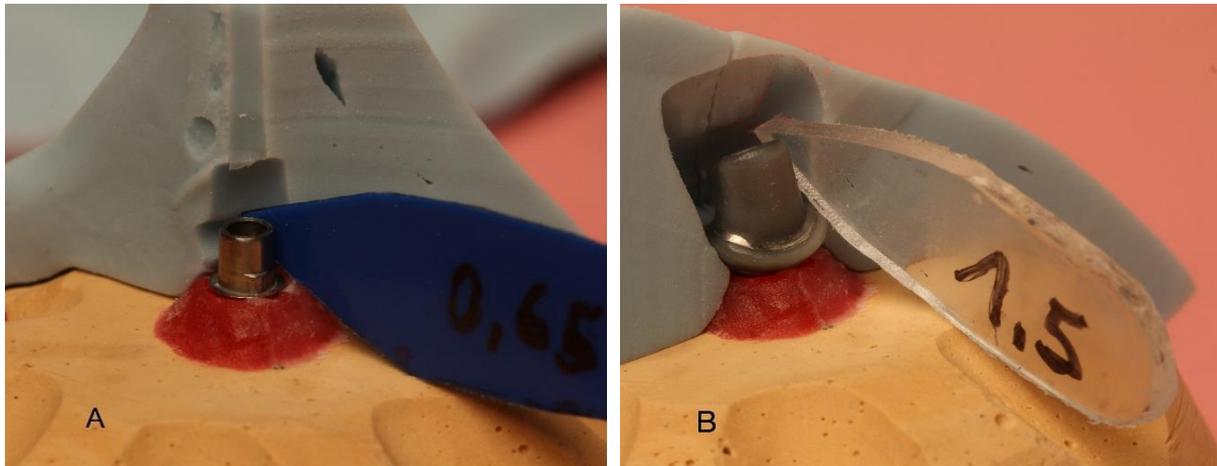


Figure 4: A silicone index was fabricated for the abutment suprastructure (A) and the crown (B). A rubber strip of 0.65 mm (A) or 1.5 mm (B) was used to ensure there was enough space for the ceramic.

In order to achieve identical dimensions during preparation of the abutments with the copy-milling technique, a standard abutment (FairTwo abutment, FairImplant) was modified according to the desired design. Then the modified abutment was scanned and all abutments of the study were milled to the same dimensions and shape for the purpose of standardization. For the fabrication of the eighty crowns, a full wax-up was done for one crown that was scanned. Then eighty crowns were milled from IPS emax CAD blocks. After receiving the abutments and crowns from the laboratory, their thicknesses were verified with the use of a caliper.

3.2.3. Assembly of the implants and abutments

Implants were embedded in a three-component, self-curing polyester resin (Technovit 4000, Heraeus Kulzer, Wehrheim, Germany) to imitate the elastic reaction of the surrounding bone during loading. The resin covered the implant body up to the first thread. Brass tubes were used as molds and also served as the specimen holders during testing.

All abutments were attached to the implants with titanium screws (FairImplant) of 9 mm length and 1 mm diameter. A new screw was used for each assembly to avoid any stress of the screw done during forehand tightening and loosening. Before the new screw was placed, antiseptic gel (Chlorhexamed, GSK, Bühl, Germany) was placed at the implant connection to simulate the clinical situation (Fig. 5). At first torque wrenches were checked with a calibrated Torque Tester (Crane Electronics, Hinckley, UK) (Fig. 6) to make sure the chosen torque was delivered during tightening. Then the screws were tightened with 25 Ncm; after 10 minutes the screws were retightened to avoid any screw loosening ^[44, 117, 118]. The screw cavities were filled with foam pellets and gutta-percha.



Figure 5: Applying antiseptic gel inside the internal connection of the implant before screwing the abutment finally to imitate the clinical situation.

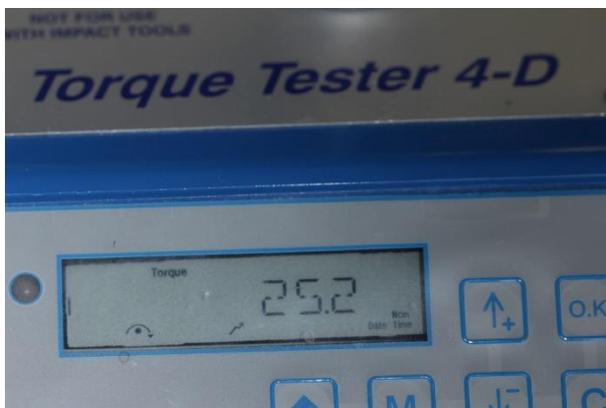


Figure 6: A torque tester was used every time before tightening the screw to calibrate the torque wrenches and ensure uniform delivery of the torque to the screws of all specimens.

3.2.4. Bonding procedure

3.2.4.1. Bonding the titanium inserts to the ceramic suprastructures

Titanium inserts (FairImplant) were adhesively cemented to the ceramic suprastructures (zirconia in group **ZrT** and lithium disilicate in groups **LaT** and **LcT**). Surface treatment of titanium inserts was done by air-abrasion using 50 µm alumina particles at 2.5 bar. The zirconia suprastructures were air-abraded using 50 µm alumina particles at 1 bar. For the lithium disilicate abutments and hybrid-abutment crowns, the inner surfaces were etched according to the manufacturers' instructions for 20 seconds with 4.5% hydrofluoric acid (IPS Ceramic Etching Gel, Ivoclar Vivadent). Cementation was done using a self-curing luting composite (Multilink Hybrid Abutment, Ivoclar Vivadent) under a constant load of 750 grams, after the surfaces were primed with a universal primer for ceramics and metals ^[16] (Monobond Plus, Ivoclar Vivadent). Excess cement was removed and a glycerin gel (Liquid strip, Ivoclar Vivadent) was applied to prevent the formation of an oxygen-inhibited layer (Fig. 7).



Figure 7: The ceramic suprastructure was bonded to the titanium insert under constant load of 750 grams to ensure uniform bonding technique for all specimens. As the load was applied, the excess cement was removed and glycerin gel was applied.

3.2.4.2. Bonding of the crowns to the abutments

For adhesive cementation of the crowns to the abutments, the bonding surfaces of titanium and zirconia abutments were air-abraded with 50 μm alumina particles at 2.5 bar pressure for titanium and 1 bar pressure for zirconia until a black marker coating was completely removed. After air-abrasion the abutments were ultrasonically cleaned in 99% isopropanol for 3 minutes and dried. The lithium disilicate abutments (group **LaT**) were etched using 4.5% hydrofluoric acid (IPS Ceramic Etching Gel, Ivoclar Vivadent) for 20 seconds. The inner surfaces of lithium disilicate ceramic crowns were also etched according to the manufacturer's instructions for 20 seconds with the 4.5% hydrofluoric acid. The external surface of the lithium disilicate abutments and the internal surface of each crown were then carefully cleaned with water spray and air dried. Bonding areas of all abutments and crowns were primed (Monobond Plus, Ivoclar Vivadent) for 60 seconds and were again air dried. The crowns were then bonded to the abutments using a dual-curing adhesive resin cement (Multilink Automix, Ivoclar Vivadent) under a constant load of 49 N (Fig. 8). After excess cement was removed, a glycerin gel (Liquid strip, Ivoclar Vivadent) was applied to the abutment-crown interface. Light curing (Elipar 2500, 3M ESPE, Neuss, Germany) was then applied for 20 seconds from labial and palatal sides.



Figure 8: The crowns were cemented under constant load of 49 N to ensure uniform bonding technique of all specimens. Glycerin gel was applied to the interface to ensure complete polymerization of the cement.

In group **LcT**, the screw channel was etched with 4.5% hydrofluoric acid (IPS Ceramic Etching Gel, Ivoclar Vivadent), silanated (Monobond Plus, Ivoclar Vivadent) and sealed with composite resin (Tetric EvoCeram, Ivoclar Vivadent). Composite resin was applied in increments and each was light cured for 20 seconds. Finishing and polishing was then done using special composite polishing set (Compositopolitur, Komet, Lemgo, Germany).

Figure 9 shows, as an example, group **LaT** after completing the bonding procedure.



Figure 9: Eight complete specimens of group LaT after bonding and before starting the tests.

3.2.4.3. Storage

The restorations were stored in distilled water at 37°C for 72 h before testing, to ensure that autopolymerization of the resin cement was complete.

3.3. Tests and statistical analysis

3.3.1. Dynamic loading test

According to the study outline (Table 3) groups **Ti1**, **Zr1**, **ZrT1**, **LaT1**, and **LcT1** (n=8) were subjected to dynamic loading in a computer-controlled dual-axis chewing simulator (Chewing Simulator CS-4, SD-Mechatronik, Westerham, Germany) for 1,200,000 loading cycles (Fig. 10) that corresponds to a 5-year clinical fatigue [67]. A loading force of 49 N was applied at an angle of 30° degrees to the implant axis, 3 mm below the incisal edge on the oral aspect of the crown at a frequency of 1.6 Hz using a ceramic ball with a 6-mm diameter (Steatite Hoechst Ceram Tec, Wunsiedel, Germany). In order to simulate wet conditions of the oral cavity and to subject the ceramic to a wet environment, all specimens were soaked in distilled water at room temperature for the whole period of testing. Video recording cameras were placed for each specimen throughout the test to detect the number of cycles the specimen survived in case of failure during dynamic loading.



Figure 10: Test specimens during dynamic fatigue loading in the chewing simulator.

3.3.2. Quasi-static loading test

All specimens of groups **Ti1**, **Zr1**, **ZrT1**, **LaT1** and **LcT1** were checked for incipient fracture or screw loosening. Then all survived specimens and all specimens of groups **Ti2**, **Zr2**, **ZrT2**, **LaT2** and **LcT2** were subjected to quasi-static loading using a universal testing instrument (Zwick Z010, Zwick, Ulm, Germany). A semi-spherical loading stamp was positioned 3 mm below the incisal edge on the oral aspect of the crown. However, a 0.5 mm-thick tin foil (Zinnfolie, Dentaureum, Ispringen, Germany) was placed between loading stamp and crown to achieve homogenous stress distribution. Then, a compressive force was applied at the same angle of 30° degrees to the implant axis (Fig. 11) under stroke control with a cross-head speed of 2 mm/min until failure which was perceived as a fracture, a sudden reduction in force or deflection of 3 mm. A video recordings of all tests were done with an integrated video camera that allows replay of the test simultaneously while checking the graph, this helps to exactly detect the force at which failure happened as well as the mode of failure. The failure loads were recorded by a commercial software (testXpert II V3.3, Zwick, Ulm, Germany).



Figure 11: A specimen of group Ti during applying quasi-static load.

3.3.3. Microscopic evaluation

After the quasi-static and the dynamic loading tests, all specimens were examined visually and under low power (50 x) stereo-magnification with the use of an optical microscope (Carl Zeiss, Jena, Germany) and representative photographs of failed specimens were taken. The microscopic evaluation was performed to assess the mode of failure. Therefore, all tested specimens were examined for incipient fractures and the mode of failure was classified according the locations of possible fractures. Randomly selected specimens were investigated using X-ray radiographs and other specimens were cut into two vertical halves after being placed in stycast for support

3.3.4. Statistical analysis

To reveal differences in the bending behavior of metal bases, the deformation that occurred in each abutment was determined for forces equivalent to 100 N, 200 N, 300 N, and 400 N. The deflection of the specimens (in μm) for each 100 N increase in force was measured and analyzed. Normality distribution was tested using Shapiro-Wilk test which revealed that the data was not normally distributed. Therefore, statistical analyses of the test results were made with the Kruskal- Wallis ($p=.001$) test followed by multiple pair-wise comparisons of the groups using Mann-Whitney tests at $p\leq.05$. Significance levels were adjusted with the Bonferroni-Holm correction for multiple testing (Fig. 12).

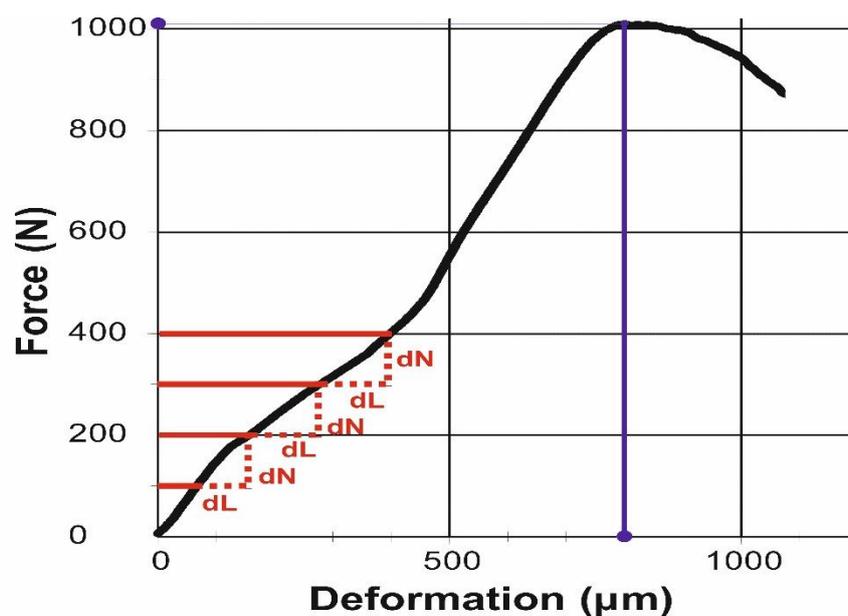


Figure 12: Typical force-deflection graph used to measure deflection (specimen of group **LcT** as an example). The red lines indicate the deflection of the specimen at 100 N, 200 N, 300 N and 400 N.

4. RESULTS

4.1. Results of dynamic loading

All test specimens of groups **Ti1**, **ZrT1**, **LaT1** and **LcT1** survived 1,200,000 cycles of dynamic loading in the chewing simulator. No screw loosening or incipient fractures in the ceramic abutments or crowns were recorded.

In group **Zr1**, three specimens failed at approximately 185.000, 230.000 and 310.000 cycles, respectively. The failure mode was similar for these three specimens and was exhibited as fracture of the zirconia abutment at the abutment-implant connection area slightly above the implant shoulder.

4.2. Results of quasi-static loading

All specimens (n=16) of groups **Ti**, **ZrT**, **LaT** and **LcT** and 13 specimens of group **Zr** (after failure of 3 specimens during dynamic loading) were subjected to quasi-static loading until fracture using a universal testing instrument (Zwick Z010, Zwick).

The median fracture load for group **Zr** was 217 N. Each subgroup showed the following median fracture loads: 198 N for subgroup **Zr1** which was subjected to dynamic fatigue loading and 218.5 for subgroup **Zr2** which was directly subjected to quasi-static loading.

When used with titanium insert, zirconia abutments (group **ZrT**) could withstand median forces up to 943 N without fracture. The lithium disilicate abutments successfully resisted fracture and tolerated median forces up to 987 N for group **LaT** and 989 N for group **LcT** (see appendices). Nevertheless, these values cannot be stated as the definitive fracture strength of the crown-abutment system as test was stopped due to plastic deformation of the titanium inserts. As plastic deformation of the abutments will be considered clinically as failure, the bending of the titanium abutments or titanium inserts was considered as failure and the test was stopped, even if the ceramic did not fracture.

4.3. Mode of failure

In group **Zr** all specimens exhibited one type of failure that was fracture of the ceramic and was located predominantly at or slightly above the level of the implant shoulder. The fractured components remained inside the internal connection part of the implants (Figs. 13 and 14).



Figure 13: A specimen of group Zr broken at the abutment neck after quasi-static loading.

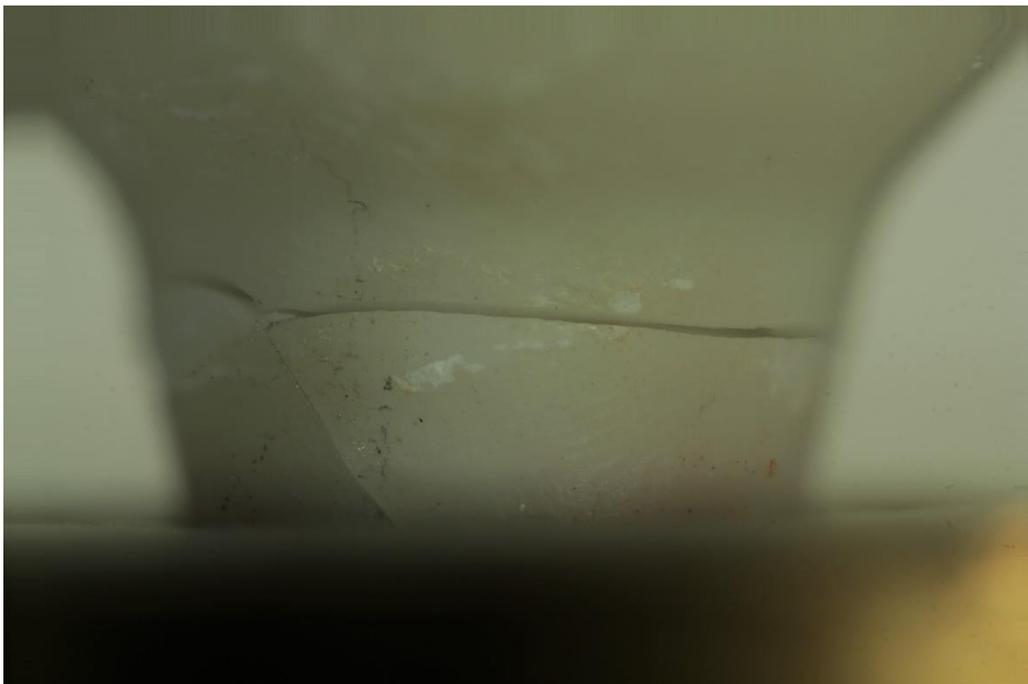
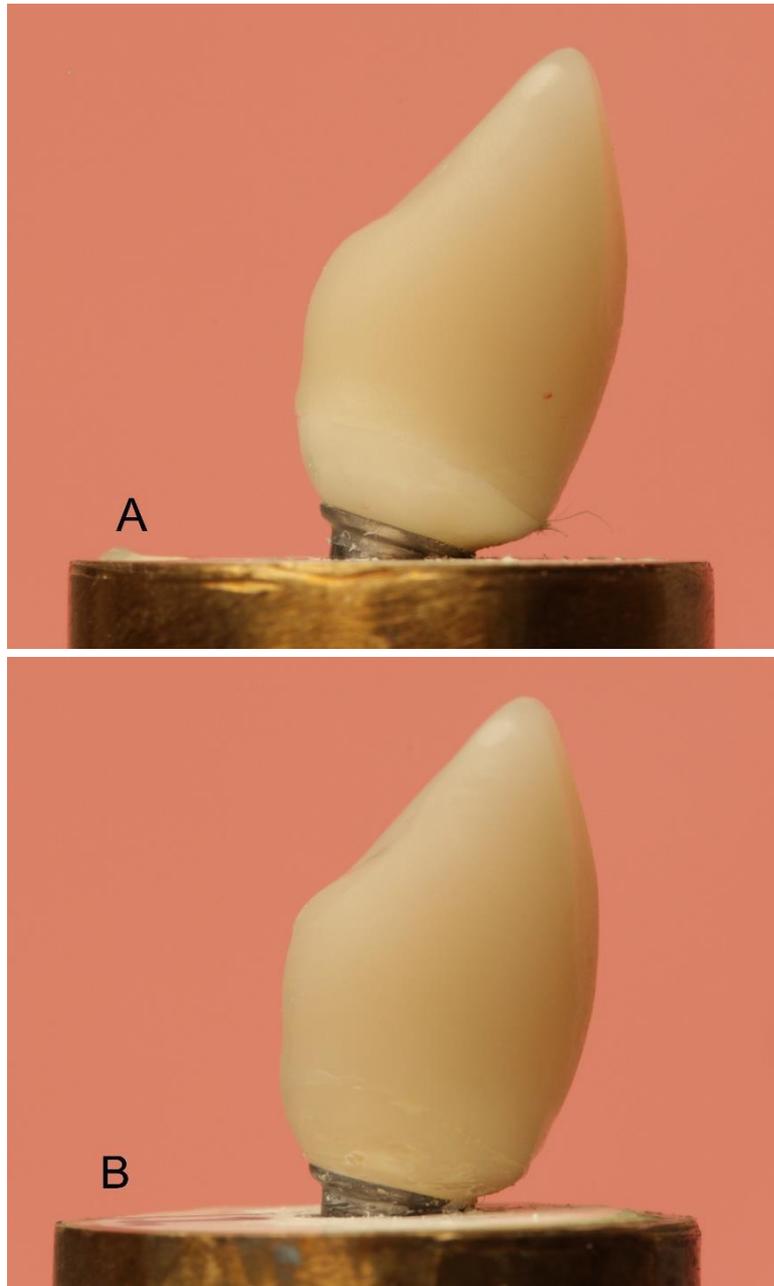


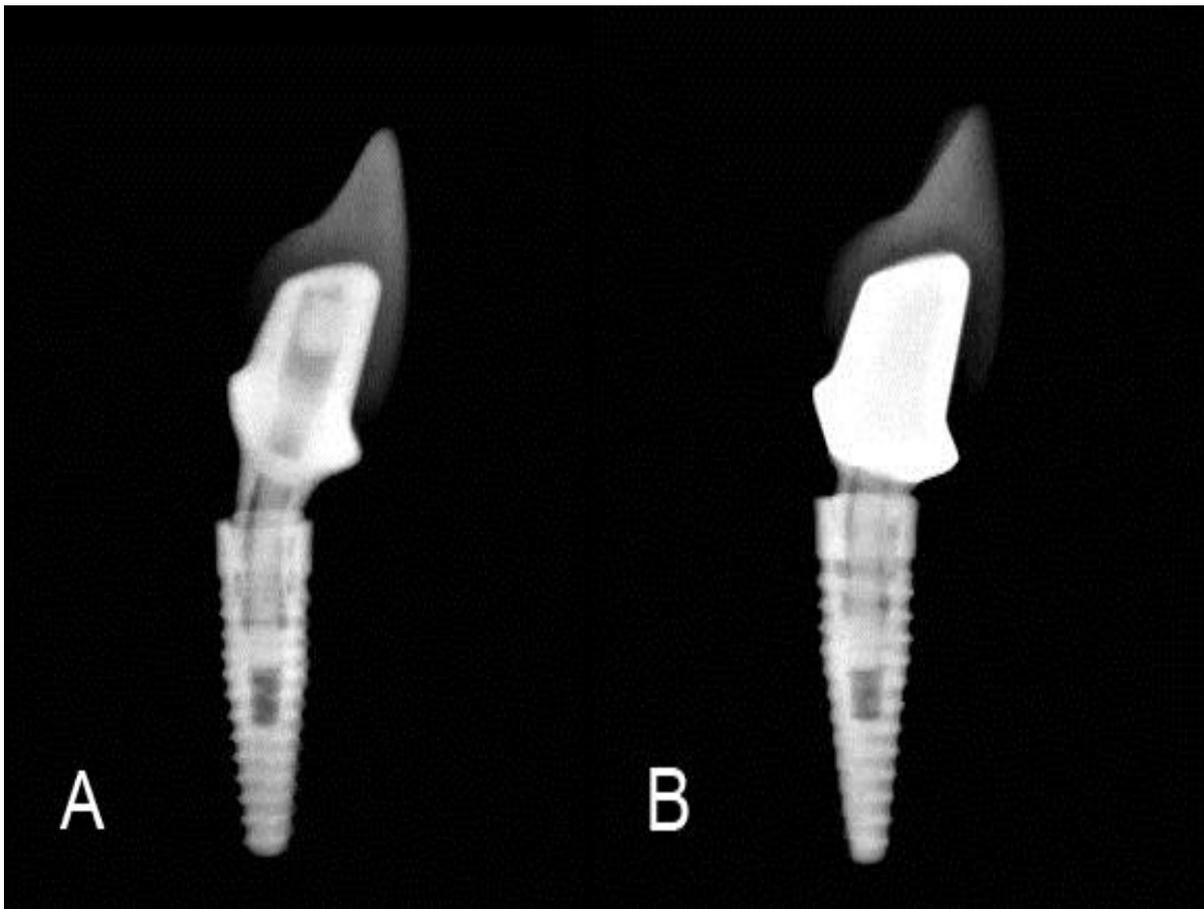
Figure 14: Microscopic picture of a failed specimen of group Zr with 40X magnification.

Visual assessment of failed specimens of the titanium abutments (group **Ti**) as well as all abutments with titanium insert (groups **ZrT**, **LaT** and **LcT**) showed homogenous mode of failure. All deflected specimens of these four groups presented permanent bending at the screw and internal connection of the titanium abutment or insert without ceramic displacement or fracture, and slight distortion of the labial implant platform (Figs. 15 A and B).



Figures 15 A and B: Failed specimens of groups ZrT (A) and LaT (B) after quasi-static loading. The specimens show plastic deformation of the titanium inserts without fracture or debonding of the ceramic suprastructure.

None of the abutments in the four groups fractured or showed mobility after testing. However, radiographic pictures of randomly selected specimens were made to check whether there was a fracture in the screw or the metal (Figs. 16 A and B). The radiographs showed bending in the screw as well as the titanium connection (abutment or insert) but no fracture. To take a closer look of what exactly happened in the implant-abutment assembly, randomly selected specimens were cut into two vertical halves (Fig. 17). The same mode of failure was also verified in these specimens.



Figures 16 A and B: Radiograph of specimen of group Ti (A) and ZrT (B) showing bending of the screw and abutment connection without fracture.



Figure 17: A sectioned specimen of group Ti, showing plastic deformation of the screw and the abutment without any fracture.

4.4. Deformation of metal

To reveal the difference of the bending behavior of metal in the four groups, the graphs produced by testXpert software of Zwick were analyzed and the forces of 100 N, 200 N, 300 N and 400 N were determined and marked on the graph. Then the deformation (in μm) of the metal done at each given force was also determined. The distance traveled by the metal (in μm) for each 100 N increase in force was measured and a table was made to show deformation from 100 N to 200 N, from 200 N to 300 N and from 300 N to 400 N for each group. The data was then analyzed using Kruskal-Wallis test followed by multiple pair-wise comparisons of the groups using Mann-Whitney tests at $p \leq 0.05$. Significance levels were adjusted with the Bonferroni-Holm correction for multiple testing to reveal statistical differences between groups (Table 4; compare to Figure 12).

The titanium abutments showed more bending than titanium inserts at a given force. However, the titanium inserts in groups **ZrT**, **LaT** and **LcT** did not show any significantly different behavior of deformation.

Table 4: Medians of traverse distance (in μm) of deformation of metal at given forces and the statistical differences.

	dL (100 - 200 N)	dL (200 - 300 N)	dL (300 - 400 N)
Group Ti	357 ^A	360 ^A	570 ^A
Group ZrT	150 ^B	163 ^B	197 ^B
Group LaT	216 ^B	155 ^B	205 ^B
Group LcT	184 ^B	163 ^B	189 ^B

*In a column, different letters indicate statistically significant differences between groups. (Mann-Whitney tests with $\alpha=.05$). Overall Kruskal-Wallis test $\alpha=.001$.

5. DISCUSSION

5.1. Discussion of methodology

To minimize the number of variables, the study was designed to limit the variables solely to the abutment materials using always an identical crown dimension and material.

5.1.1. Implants

An internal connection platform-switched implant with a diameter of 4.2 mm was used. This diameter complies with anatomic constraints that may present in the anterior esthetic region preventing a larger implant diameter to be inserted ^[45]. Moreover, an internal friction connection connects an implant with an abutment by means of screw tightening and friction occurring at the contact between the implant and the abutment. This internal friction connection is known to provide more intimate contact with implants than an external connection ^[91].

5.1.2. Screw tightening

In this study, implants were embedded in an autopolymerizing acrylic resin to imitate the human bone as this may have a better stress-distribution effect ^[13].

Before tightening the screws, a calibrated torque tester was used to check whether the torque indicated on the torque wrench was delivered to the screw. During this procedure it was shown that the torques stated on the torque wrench were not precise, thus it needed calibration to deliver the required torque values, which proved the importance of the calibration step before tightening the screws. During laboratory adjustment and surface treatment of the abutments and titanium inserts, they were assembled on laboratory analogues which required repetitive screw tightening and loosening. This might have led to stresses induced in the screws and could cause loosening or fracture during loading ^[26]. Therefore, a new screw was used every time before final assembly of the abutments on the implants.

Farina et al. tested different tightening protocols for titanium and gold screws and found that application of a retorque after 10 minutes of the initial torquing increased joint stability independent of fit level or screw material ^[44]. Siamos et al. also recommended that retightening abutment screws 10 minutes after the initial torque applications should be routinely performed, after applying different torquing protocols and subjecting specimens to cyclic loading ^[117].

This is due to embedment relaxation (settling), a mechanical engineering principle that affects preload. Because the internal threads of the implant and the screw threads that contact these internal threads cannot be machined perfectly smooth, high spots will inevitably occur on both surfaces. These high spots will be the only contacting surfaces when the initial tightening torque is applied to the screw and the preload is developed. Embedment relaxation then occurs, whereby the rough spots actually flatten (or wear) under loading, and 2% to 10% of the initial preload is lost ^[61]. After embedment relaxation, applying a tightening torque will once again act to regain preload. Therefore, the screws were tightened at 25 Ncm using the calibrated torque wrenches and then after 10 minutes they were tightened again using the same torque.

5.1.3. Titanium inserts

The original titanium inserts provided by the implant manufacturer (FairImplant) was 5 mm in height. This was also the minimum height of titanium inserts required to bond IPS emax CAD according to the manufacturer's instructions (Ivoclar Vivadent). So after personal discussion with the both companies' research advisors (FairImplant and Ivoclar Vivadent), a reduction in the height of the titanium inserts was not suggested by both companies as this may cause weakening of the ceramic abutments. However, when designing the abutments and crowns, it was found that if the original titanium abutments and titanium inserts were to be used, there will be a space of only 0.38 mm for the ceramic material circumferentially around the insert, whereas the minimum layer thickness required for IPS emax CAD abutments is 0.5 mm (Fig. 18). Therefore, a new design for both the inserts and the abutments was needed. In addition, small spaces may be available clinically to restore missing teeth, e.g. maxillary lateral incisors or mandibular incisors, making it difficult to use a 5 mm titanium base and on top of it an abutment and a crown. Moreover, even if the space is available, such a long titanium insert can hinder esthetics which is the main reason for using a ceramic abutment. So the decision was made to reduce the height of the titanium inserts to 3 mm, not following the instructions of the manufacturers but rather simulating a clinical situation that is likely to happen.

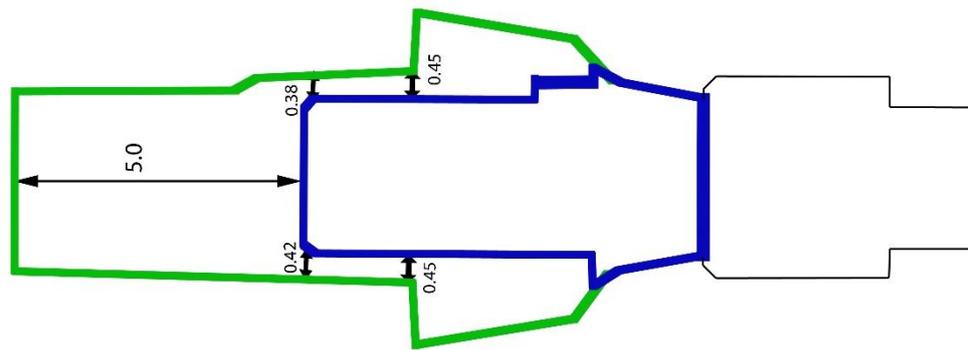


Figure 18: A schematic drawing of the original titanium base (5 mm) (blue) and a ceramic suprastructure (green) with the dimensions of the original abutment. As shown, a minimum thickness of 0.5 mm would not have been achieved if this design had been used.

5.1.4. Abutments

After reduction of the titanium base to 3 mm, preparation of the original titanium abutment (FairImplant) was done in respect to the manufacturer's preparation guidelines for the minimum thickness of 0.5 mm required for the zirconia and lithium disilicate abutments in combination with lithium disilicate crowns. In several clinical and laboratory studies, circumferential shoulder preparations of 1.0 to 1.5 mm were routinely used for the fabrication of all-ceramic crowns [14, 21, 106]. Lin et al. found that the preparation finish line can affect the marginal adaptation of the crown [81]. Using the chamfer preparation in comparison to the shoulder one seems to improve the marginal fit [101]. It may also help adhesive cement, used in this study, to escape during seating and therefore to improve marginal adaptation [46]. Therefore, the modified abutment was prepared with a circumferential chamfer preparation. However, sharp transitions and inner angles were avoided.

Then the modified abutment was scanned. Milling of all titanium and zirconia abutments as well as the zirconia and lithium disilicate suprastructures was done according to the scanned modified abutment using CAD/CAM technology. Milling the zirconia in its soft state and then sintering it in its final shape provides the maximum amount of tetragonal phase (Y-TZP), which is an advantage of CAD/CAM fabrication [85]. Besides, using a custom-made design allows manufacturing of ceramic abutments optimized to the anatomic dimensions and soft tissue contour. This provides a natural-looking emergence profile to harmonize the restoration with

the adjacent natural teeth. Due to the anatomical marginal configuration of the abutments there was no need for preparing mechanical anti-rotational features.

The cervical region of the abutments represents the area of the highest torque and stress concentrations, therefore, the implementation of a titanium insert is important to replace the brittle ceramic with metal. This is why the lithium disilicate abutments were exclusively assembled with a titanium insert. This also complies with the manufacturer's instruction.

5.1.5. Crowns

In general the thickness of anterior all-ceramic crowns may differ from the incisal edge to the axial surfaces and is strongly influenced by the preparation design. A minimum reduction of 1.5-2 mm in height and 1.0 to 1.5 mm circumferentially is critical for both the stability of anterior all-ceramic crowns under functional loading and their esthetic performance.

A full wax-up of a right maxillary central incisor was done on the modified abutment. The crown gingivo-incisal length was 11.0 mm and the mesio-distal width at the incisal edge was 8.5 mm. For means of standardization, using CAD/CAM technology the wax-up of the crown was scanned and then eighty crowns were milled from IPS emax CAD blocks.

As recommended for lithium disilicate crowns, a 6° convergence chamfer preparation with rounded inner angles was designed to provide adequate mechanical retention ^[15, 21] and to improve marginal fit adaptation ^[101, 116].

5.1.6. Cementation

Etching and silanization seem to be not effective in the case of zirconia, since it presents a very dense morphology which contains no glass phase. Similarly, it has been reported that silica coating provides a non-durable bond to Y-TZP ^[68]. Only bonding systems that contain a special adhesive monomer have been found to provide an acceptable, high strength, durable bond to air-abraded Y-TZP. However; the retention can be increased for instance by air-abrasion with 50 µm alumina particles ^[124]. So in this study air-abrasion of zirconia abutments and suprastructures was done before cementation to either titanium inserts or lithium disilicate crowns.

Ebert et al. tested different methods of surface treatment before bonding zirconia suprastructure to titanium inserts. After the specimens had been stressed for either 1, 30, 60, or 150 days by water and thermal cycling, retention was measured, and it was found that air

abrasion increased the retention significantly ^[41]. Therefore, this method was followed in this study.

On the other hand, bonding to silica based ceramic may be very effective by using a resin luting agent after hydrofluoric acid etching, which can create a micro-retention pattern on the ceramic surface by dissolving silicate components, and silanization, which forms a chemical link to the glass-ceramic surface and provides better wetting ^[23, 72]. These guidelines, which comply with the manufacturer's instructions, were followed while bonding the lithium disilicate to the titanium base, as well as bonding the lithium disilicate crowns to all abutments. The choice of the cements was done to follow the company's recommendation. The adhesive cementation of lithium disilicate crowns over the prepared and air-abraded part of the zirconia abutments may enhance the fracture strength because it seals the prepared surface and potential superficial micro cracks.

5.1.7. Tests

Before materials should be used clinically, scientific data should be available to support the mechanical and biological properties of the material. Clinical studies of at least five years are recommended, as materials and restorations are more likely to fail after five years or more. However, clinical studies that could accurately evaluate the biomechanical behavior or the clinical success of materials and restorations need increased costs and time ^[67]. Therefore, a well-designed in-vitro study that simulate clinical conditions can be an effective indicator of the performance of the materials and its likely clinical success, before widespread clinical use is recommended.

Several factors influence the fracture loads of all-ceramic crowns, such as the microstructure of the ceramic material ^[36, 98], the fabrication technique, the final surface finishing of the crowns and the luting method ^[24, 88]. Several factors influence the fracture loads of all-ceramic including test conditions such as the storage conditions, the type of the fatigue test used, and the direction or/and the location of the loading force ^[65, 128]. The study was designed to standardize all these factors as close as possible to the clinical conditions.

The direction of the loading forces may significantly influence the fracture strength of all ceramic restorations ^[74]. In this study, force application at an angle of 30 degrees was chosen during dynamic as well as static loadings, to represent a clinical occlusal force application ^[51, 54, 99] forming an interincisal angle of 150 degrees in a Class I occlusion ^[8].

The clinical failures of dental restorations most commonly result from fatigue ^[11]. Cyclic loading has been demonstrated to decrease the fracture resistance of ceramic abutments as well. Gehrke et al reported a decrease in the strength of zirconia abutments from 672 N to 405 N after cyclic loading ^[47]. Thus, to see the effect of the fatigue loading on the abutments, the study design was made to artificially age half of the specimens by applying dynamic loading with parameters similar to those reported in the literature ^[12, 97].

The main objective of testing in a chewing simulator was to introduce a comparable cycle fatigue component to that found physiologically in the oral cavity. Chewing consists a high number of low cyclic loads, therefore fatigue loading in a chewing simulator that generates cyclic patterns with physiological load characteristics could be gathered more as clinical relevant testing conditions than monotonic loading ^[65]. The parameters used for the chewing simulation in this study were designed to approach a clinically relevant situation. The effective weight of each antagonistic specimen was 5 kg, which corresponds to a loading force of 49 N. Several in vitro studies used a cycle loading force of 49 N for chewing simulation ^[13, 25, 67, 97, 119]. These studies have considered the functional forces that arise during mastication or swallowing, which usually range between 2 and 50 N ^[19, 35, 75, 76, 119]. It was shown that humans have an average of 250,000 masticatory cycles per year ^[37, 38, 112]. Therefore, to simulate a clinical service time of 5 years, a total of 1,200,000 masticatory cycles should be performed during chewing simulation ^[67, 119].

In addition, the specific chewing simulator used in this study was developed to reproduce testing conditions under controlled moisture. Exposure to water has been found to induce aging-related phenomena which result in surface degradation of the material and thus affect the mechanical properties of zirconia ceramics ^[31]. Steatite ceramic showed a similar hardness to enamel (Vicker's scale) and was gathered as a suitable substitute material for enamel in wear tests ^[65], therefore it was used in this study as antagonist during dynamic loading.

5.2. Discussion of results

5.2.1. Dynamic loading

In the present study, each test group was divided into two subgroups and one of them was exposed to the chewing simulator before the fracture strength test was performed. During the dynamic loading of the five subgroups, all specimens of subgroups **Ti1**, **ZrT1**, **LaT1** and **LcT1**

survived 1,200,000 cycles of exposure to the simulated oral environmental testing. In group **Zr1**, five specimens survived the dynamic fatigue loading test while three specimens showed fracture of the zirconia during the test. These three specimens survived only 185,000, 230,000 and 310,000 loading cycles, respectively. Before surviving specimens of all groups were tested with quasi-static loading, they were checked visually and under a microscope to detect any incipient fractures or screw loosening. But all components of the specimen, including the implant, abutment, screw and the all-ceramic crown, were found in perfect condition without any fractures.

As mentioned before, according to the literature, an average of approximately 250,000 cycles in a chewing simulator corresponds to one year of clinical service. Therefore, the 1,200,000 dynamic loading cycles, achieved by the all-ceramic implant restorations in the three study groups and five specimens of group **Zr1**, without failure, corresponded to a 5-year service time.

Laboratory studies support the use of zirconia abutments in the anterior region after exploring the feasibility to withstand functional loading in the oral environment [12, 13, 25]. In these studies, single implant all-ceramic crowns of a maxillary incisor placed on zirconia abutments were tested up to 1,250,000 cycles in a chewing simulator under a loading force of 30 to 49 N. The aforementioned restorations in these studies noted survival rates of 100% after an equivalent of 5-year chewing simulation without any screw loosening in agreement to the present study. This could be also verified by a recent systematic review that showed a clinical failure rate of ceramic abutments of 2.5% after 5 years [131].

5.2.2. Quasi-static loading

Maximal occlusal forces reported in the anterior region were in the range of 150–235 N with a mean of 206 N [54]. Bruxism and other functional disorders can induce higher bite forces [96].

Loads of these magnitudes were tolerated by specimens of groups **Ti**, **ZrT**, **LaT** and **LcT** but not by specimens from group **Zr**. The pure zirconia abutments in this study had a median fracture resistance of 217 N. This value is considered to be in the range of the physiological maximum occlusal forces in the anterior region, indicating a risk of failure when using this abutment type clinically.

Obviously, the application of a secondary titanium insert positively influences the performance of zirconia abutments through replacing the brittle zirconia with titanium at the implant-abutment connection area. Likewise, the present study showed that the internal friction connection that used a titanium insert with a zirconia abutment could withstand median forces up to 943 N without fracture, making it much more resistant to loading than the pure zirconia abutment. This result corresponds to that of previous studies [33, 122, 127]. Furthermore, the difference between the fracture mode of the **Zr** and **ZrT** groups was prominent.

In this study, titanium, hybrid zirconia and both types of hybrid lithium disilicate abutments could bear loads more than the reported physiological maximum occlusal forces in the anterior regions. Although deformed, hybrid zirconia and lithium disilicate abutments successfully resisted fracture and could tolerate forces up to range of 900 N. Yet, these values cannot be stated as the definitive fracture strength of the materials as there was reduction of force and test was stopped due to plastic deformation of the titanium inserts which was deemed as failure. Thus, all test specimens of these groups exceeded the minimum limits of the fracture resistance for anterior restorations.

5.2.3. Failure mode

Study findings in all of the specimens of the pure zirconia abutments (group **Zr**) revealed that fractures were located at the cervical aspect of the abutments at or slightly above the level of the implant-abutment internal connection. Fractures occurred through the most tapered part, towards the platform level and this typical failure pattern was observed in all groups regardless the loading mode. No damage or plastic deformation of the implant or abutment screw happened. This is consistent with results of other studies [13, 45, 94, 97, 126].

Without exceptions, all specimens of test groups **Ti**, **ZrT**, **LaT** and **LcT** (with titanium abutments or titanium inserts) showed a uniform failure mode which was plastic deformation of the titanium. Failure of zirconia or lithium disilicate due to fracture or debonding between the metal and ceramic did not occur in the current study.

The mean fracture strength of the lithium disilicate ceramics is lower than that of zirconia [5, 57, 86]; yet, statistical analysis of this study revealed that there was no significant difference between the use of both materials as hybrid abutments. This is due to the fact that the failure occurred as a result of deformation of the titanium inserts and screws which were similar in both types of abutments, whereas the ceramic suprastructure remained intact. The aim of this

study was to test the failure of the abutments in respect to clinical situation, hence, testing was stopped after 3 mm of deformation of the abutments, even if ceramic fracture did not occur. However, if the loading tests were continued, difference in the fracture resistance between zirconia and lithium disilicate may have been noticed. Similarly, no differences between abutments and hybrid-abutment-crown made from lithium disilicate were observed, as again, the failure occurred at the titanium inserts and not at the ceramic suprastructure. It can be assumed that the adhesive cementation strengthened the connection between the abutment and the crown. Abutment screw deformation would not be a concern in clinical situations because the screws can withstand typical human occlusal forces ^[70].

During screw tightening, titanium allows some favorable degree of elastic deformation and accommodates the plastic deformation generated by friction between the different components ^[2]. This is known as the settling effect ^[125]. As loads increase beyond the yield limit of the titanium abutment, the components deform and bend, which may lead to the fracture of the abutment screw as it is the weakest component ^[45]. This explains the high degree of deformation of the titanium abutment and titanium inserts in this study.

The bending behavior of the titanium insert in specimens of groups **ZrT**, **LaT** and **LcT** was similar. However, the titanium abutments exhibited higher bending corresponds to the load values of the titanium inserts. According to the manufacturer's explanation, the titanium abutments used in this study were milled from blanks in a dental laboratory to produce the customized design of the abutment requested. In contrast, the titanium inserts were prefabricated in the factory with different milling parameters.

In a second class lever the input effort is located at the end of the bar and the fulcrum is located at the other end of the bar, opposite to the input, with the output load at a point in between the input and the fulcrum. When abutments with internal conical connection are used and subjected to forces applied at an angle of 30° to the implant axis, second class levering effects are induced. Therefore, the output load is applied in area of the internal cone of the abutment. Thus, internal cone of the abutment seems to be a high loaded component that receives torque and stress concentrations. This might explain why all abutments failed at the area of connection, which was seen in either fracture of zirconia in group **Zr** or metal deformation at this particular area in the other four groups. However, it was illustrated both in laboratory and in clinical studies that an internal connection of abutments tends to be beneficial regarding fracture strength of the abutment and screw stability ^[110].

5.3. Study limitations

Differences in results and mode of failures between studies can be contributed to several factors. Differences in grades of titanium and the constituents of zirconia ^[33] are possible reasons for the different fracture strengths reported in different studies. Additionally, the load bearing capacity of the implant restorative system depends on the implant-abutment connection. The performance of titanium abutments, when loaded, for systems with conical connections seems to be inferior to that of external hexagon connections ^[108, 114]. It has been suggested that a platform-switching design presents the disadvantage of higher stresses in the abutment and screw as a result of the smaller diameter of the connection in comparison to that of the external hexagon design ^[87, 109, 114]. Also zirconia abutments have different capacities to withstand forces depending on the manufacturing and handling processes and the specific connection design of different systems ^[114].

The number of specimens tested and the use of water rather than artificial saliva during testing present limitations of the current study. Another limitation of the study is related to the location of the applied load. The anterior part of the tip of the load applicator was positioned 3 mm below the incisal edge at the palatal surface. This model resembles a class I dentition. Yet, in cases of class II dentition, where the load is applied more cervically, or class III, where the load is applied more incisally, the force distribution will be different and the fracture mode and load may also be different.

6. CONCLUSIONS

Within the limits of this laboratory study, the following conclusions can be drawn:

1. Lithium disilicate abutments have the potential to withstand physiologic occlusal forces applied in the anterior region, and can therefore be recommended as an esthetic alternative for the restoration of single implants in the anterior region.
2. The fracture strength of lithium disilicate abutments is not influenced when used as hybrid-abutment or as hybrid-abutment-crown.
3. Zirconia abutments combined with titanium inserts have much higher fracture strength than pure zirconia abutments. Therefore, caution should be taken when using pure zirconia abutments.

7. SUMMARY

Recently, lithium disilicate ceramics are used to manufacture tooth-colored implant abutments due to their excellent esthetic properties. These restorations can be milled either separately as abutment and crown or in one piece as so-called hybrid-abutment-crown. The later can offer a solution to the problems which rise when cement-retained restorations are used on implants. This study was designed to measure the fracture strength and determine the failure mode of all-ceramic implant abutments and crowns fabricated from lithium disilicate and zirconia and to compare them with crowns on standard titanium abutments. In addition, the influence of long-term chewing simulation was evaluated.

Eighty implants (FairTwo, FairImplant, Germany) having a length of 11.5 mm and a diameter of 4.5 mm were used to be restored with single lithium disilicate crowns (IPS emax CAD, Ivoclar Vivadent, Liechtenstein). Five types of abutments were used; **Ti** titanium (CoproTi-4, Whitepeaks, Germany), **Zr** zirconia (Wieland Zenostar, Wieland Dental, Germany) with no titanium insert, **ZrT** zirconia with titanium insert, **LaT** lithium disilicate (IPS emax CAD, Ivoclar Vivadent, Liechtenstein) abutments and **LcT** lithium disilicate hybrid-abutment-crowns. Sixteen specimens of each group were prepared and subdivided into 2 subgroups (n=8). One subgroup (1) of each group was subjected to dynamic fatigue loading for 1,200,000 cycles at 30° to the implant axis and using load of 49 N in a chewing simulator (Chewing Simulator CS-4, SD-Mechatronik, Westerham, Germany), followed by quasi-static loading of the specimens that had survived the dynamic loading cycles. The second subgroup (2) of each group was quasi-statically loaded directly at 30° to the implant axis in a universal testing machine (Zwick Z010, Zwick, Germany) until failure of the restoration occurred.

Three specimens of group **Zr** failed to survive the dynamic loading cycles, as they were fractured at 185,000, 230,000 and 310,000 loading cycles, respectively. All other specimens survived 1,200,000 cycles of dynamic loading. The median fracture load for **Zr** was 217 N. All specimens exhibited one type of failure that was fracture of the ceramic and was located predominantly at or slightly above the level of the implant shoulder.

All specimens of the titanium abutments (group **Ti**) as well as all abutments with titanium inserts (groups **ZrT**, **LaT** and **LcT**) showed similar modes of failure. They presented bending at the screw and internal connection of the titanium abutment or titanium insert without ceramic displacement or fracture. There was significant difference between the titanium abutments and

the three groups with ceramic abutments and titanium inserts regarding degree of plastic deformation, as titanium abutments showed more bending than titanium inserts. However, the titanium inserts in groups **ZrT**, **LaT** and **LcT** did not show any significantly different behavior of deformation.

Pure zirconia abutments had a fracture resistance which is considered to be in the range of the physiological maximum occlusal forces in the anterior region. This indicates a risk of failure when using this type of abutment clinically that should be avoided. However, zirconia and both types of lithium disilicate abutments with titanium inserts could bear loads higher than the reported physiological maximum occlusal forces in the anterior region. These results support the initiation of clinical studies with such restorations in the anterior area.

8. ZUSAMMENFASSUNG

Aufgrund seiner hervorragenden ästhetischen Eigenschaften werden in der Zahnmedizin seit kurzem zahnfarbene Lithium-Disilikat-Abutments verwendet. Diese können entweder getrennt als Abutment und Krone oder einteilig als sogenannte Hybrid-Abutment-Krone hergestellt werden. Ziel dieser Studie war es, die Bruchfestigkeit und den Versagensmodus von vollkeramischen Kronen auf Keramik-Abutments (Lithium-Disilikat und Zirkonoxid) zu bestimmen und sie mit vollkeramischen Kronen auf Standard-Titan-Abutments zu vergleichen. Zusätzlich sollte der Einfluss einer lang andauernden Kausimulation evaluiert werden.

Achtzig Implantate (FairTwo, FairImplant, Germany) mit einer Länge von 11,5 mm und einen Durchmesser von 4,5 mm wurden verwendet, um Einzelimplantatkronen aus Lithium-Disilikat (IPS emax CAD, Ivoclar Vivadent, Liechtenstein) herzustellen. Fünf Arten von Abutments wurden verwendet; **Ti** Titan (CoproTi-4, Whitepeaks, Germany), **Zr** Zirkonoxid (Wieland Zenostar, Wieland Dental, Deutschland) ohne Titanbasis, **ZrT** Zirkonoxid mit Titanbasis, **LaT** Lithium-Disilikat (IPS emax CAD, Ivoclar Vivadent, Liechtenstein) Abutments und **LcT** Lithium-Disilikat Hybrid-Abutment-Kronen. Jede Gruppe hatte sechzehn Proben, die in zwei Untergruppen geteilt wurden. Proben in der ersten Untergruppe (1) wurden dynamisch in einem Kausimulator (Chewing Simulator CS-4, SD-Mechatronik, Deutschland) belastet. Belastungskräfte von 49 N wurden in einem Winkel von 30 Grad zur Implantatachse bis zu 1.200.000 Zyklen aufgebracht. In der zweiten Untergruppe (2) wurden die Proben direkt quasi-statisch, in einem Winkel von 30° zur Implantatachse bis zum Versagen der Abutments, belastet (Zwick Z010, Zwick, Deutschland).

In der Gruppe **Zr** versagten drei Proben während der dynamischen Belastung, indem sie bei 185.000, 230.000 und 310.000 Zyklen frakturierten. Alle anderen Proben überstanden 1.200.000 Zyklen dynamischer Belastung. Der Median der Bruchfestigkeit der Gruppe **Zr** betrug 217 N. Alle Proben dieser Gruppe zeigten die gleiche Versagensart, da alle Proben im Verbindungsbereich zu den Implantaten frakturierten.

Alle Proben mit Titan-Abutments (Gruppe **Ti**) sowie alle Abutments mit Titanbasen (Gruppen **ZrT**, **LaT** und **LcT**) wurden durch eine plastische Verformung des Titans zerstört. Ein Bruch oder ein Ablösen der Basis von der Mesostruktur trat nicht auf. Die Verformung war bei der Gruppe **Ti** signifikant größer als bei den Gruppen, die mit Keramik-Abutments mit

Titanbasis versehen waren. Bei den Gruppen mit einer Titanbasis (Gruppen **ZrT**, **LaT** und **LcT**) gab es hinsichtlich der plastischen Deformation keine signifikanten Unterschiede.

Reine Zirkonoxid Keramik-Abutments wiesen eine Bruchfestigkeit auf, die im Bereich der physiologischer Weise im Frontzahnbereich auftretenden Kaukräfte lagen. Ihr klinischer Einsatz würde daher ein Frakturrisiko beinhalten, das vermieden werden sollte. Hingegen wiesen Zirkonoxid-Abutments und beide Arten von Lithium-Disilikat-Abutments jeweils mit Titaninserts eine höhere Belastbarkeit auf als die im Frontzahnbereich auftretenden physiologischen Kaukräfte. Die vorliegenden Ergebnisse unterstützen daher die Initiierung klinischer Studien mit derartigen Restaurationen im Frontzahnbereich.

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11. Appendix

Table 5: Fracture strengths (N) of all specimens of group Zr				
Subgroup	Specimen no.	Fatigue Test (Survival)	Fracture Strength (N)	Fracture mode
1 (Fatigue load)	1	Failed (185.000)	-	at implant shoulder
	3	Survived	126	above implant shoulder
	4	Survived	243	above implant shoulder
	8	Survived	89	above implant shoulder
	10	Survived	257	above implant shoulder
	11	Failed (310.000)	-	above implant shoulder
	14	Survived	198	above implant shoulder
	15	Failed (230.000)	-	above implant shoulder
Median			198	
Mean			183	
SD			73	
SD (percentage)			59%	
2 (Quasi-static load only)	2	N/A	246	above implant shoulder
	5		199	above implant shoulder
	6		216	above implant shoulder
	7		216	above implant shoulder
	9		237	above implant shoulder
	12		217	above implant shoulder
	13		220	above implant shoulder
	16		226	above implant shoulder
Median			219	
Mean			222	
SD			14	
SD (percentage)			7%	

Table 6: Maximum force (N) recorded by specimens of group Ti			
Subgroup	Specimen no.	Fatigue Test (Survival)	Fracture Strength (N)
1 (Fatigue load)	2	Survived	534
	5	Survived	319
	7	Survived	502
	10	Survived	585
	12	Survived	612
	13	Survived	527
	14	Survived	511
	16	Survived	557
Median			531
Mean			518
SD			89
SD (percentage)			17%
2 (Quasi-static load only)	1	N/A	626
	3		421
	4		604
	6		538
	8		335
	9		567
	11		524
	15		692
Median			553
Mean			538
SD			115
SD (percentage)			21%

Table 7: Maximum force (N) recorded by specimens of group ZrT			
Subgroup	Specimen no.	Fatigue Test (Survival)	Fracture Strength (N)
1 (Fatigue load)	1	Survived	1000
	2	Survived	896
	3	Survived	949
	4	Survived	938
	5	Survived	888
	6	Survived	1080
	9	Survived	800
	11	Survived	963
Median			944
Mean			939
SD			86
SD (percentage)			9%
2 (Quasi-static load only)	7	N/A	940
	8		935
	10		878
	12		943
	13		943
	14		1020
	15		952
	16		975
Median			943
Mean			948
SD			40
SD (percentage)			4%

Table 8: Maximum force (N) recorded by specimens of group LaT			
Subgroup	Specimen no.	Fatigue Test (Survival)	Fracture Strength (N)
1 (Fatigue load)	2	Survived	1100
	4	Survived	1020
	5	Survived	988
	6	Survived	798
	8	Survived	986
	9	Survived	903
	10	Survived	1080
	15	Survived	1040
Median			1004
Mean			989
SD			95
SD (percentage)			10%
2 (Quasi-static load only)	1	N/A	930
	3		682
	7		933
	11		888
	12		1090
	13		998
	14		1190
	16		895
Median			932
Mean			950
SD			151
SD (percentage)			16%

Table 9: Maximum force (N) recorded by specimens of group LcT			
Subgroup	Specimen no.	Fatigue Test (Survival)	Fracture Strength (N)
1 (Fatigue load)	3	Survived	1020
	6	Survived	932
	8	Survived	1080
	9	Survived	1040
	10	Survived	980
	11	Survived	991
	13	Survived	1180
	15	Survived	908
Median			1006
Mean			1016
SD			93
SD (percentage)			9%
2 (Quasi-static load only)	1	N/A	787
	2		854
	4		988
	5		979
	7		989
	12		1010
	14		993
	16		944
Median			984
Mean			943
SD			80
SD (percentage)			9%

Table 10: Traverse distance of deformation (dL) at each 100 N for specimens of group Ti

Specimen Nr.	dL (100 N)	dl (200 N)	dl (300 N)	dl (400 N)	dL (100 - 200 N)	dL (200 - 300 N)	dL (300 - 400 N)
1	2922	3408	3749	4333	486	341	584
3	3130	3673	4519	5585	543	845	1066
4	2070	2474	2916	3476	404	442	560
5	3894	4388	4858	5498	494	470	640
12	3248	3767	4413	5334	519	646	921
14	2798	3302	4004	4896	504	702	891
16	1720	2242	2866	3719	522	624	853
2	1107	1286	1483	1808	179	197	325
7	334	507	693	935	173	186	242
10	1637	1755	1916	2023	119	161	107
12	502	643	823	925	141	180	156
13	446	614	832	988	169	218	156
14	642	780	978	1134	138	198	156
16	670	784	935	1137	113	151	202
Median	1678	1999	2391	2749	349	356	442
Mean	1913	2259	2763	2985	346	504	486
SD	1363	1546	1984	1890	201	537	71

Table 11: Traverse distance of deformation (dL) at each 100 N for specimens of group ZrT

Specimen Nr.	dL (100 N)	dl (200 N)	dl (300 N)	dl (400 N)	dL (100 - 200 N)	dL (200 - 300 N)	dL (300 - 400 N)
7	394	531	650	812	137	119	162
8	107	191	369	579	85	177	211
10	112	166	339	538	53	173	199
12	360	497	659	880	137	162	221
13	472	622	781	994	151	159	213
14	524	810	937	1118	285	127	181
15	117	271	489	697	155	218	208
16	370	508	667	864	139	159	197
7	787	927	1091	1250	139	164	159
8	939	1088	1275	1466	149	187	191
10	1375	1519	1701	1937	144	183	236
12	1255	1440	1623	1816	185	183	193
13	929	1103	1287	1483	174	184	196
14	1288	1448	1595	1792	161	147	197
15	1179	1371	1483	1653	191	112	170
16	901	1052	1232	1456	151	180	224
Median	656	868	1014	1184	150	168	197
Mean	694	846	1011	1209	152	165	197
SD	442	464	461	461	49	28	22

Table 12: Traverse distance of deformation (dL) at each 100 N for specimens of group LaT

Specimen Nr.	dL (100 N)	dL (200 N)	dL (300 N)	dL (400 N)	dL (100 - 200 N)	dL (200 - 300 N)	dL (300 - 400 N)
2	698	1111	1248	1426	412	138	178
4	646	1069	1219	1394	423	150	175
5	597	891	1044	1256	294	153	212
6	494	666	893	1114	171	227	221
8	744	912	1108	1311	168	195	204
9	362	527	736	968	165	209	233
10	665	1072	1235	1440	407	163	205
15	794	1174	1309	1488	381	135	179
1	595	768	931	1139	173	163	208
3	523	696	902	1187	173	206	285
7	580	728	862	1068	148	135	205
11	865	1154	1290	1511	289	136	221
12	525	824	961	1209	298	138	248
13	448	598	756	950	151	157	195
14	687	968	1064	1214	281	96	150
16	705	835	981	1184	130	146	203
Median	622	863	1012	1211	227	151	205
Mean	621	874	1034	1241	254	159	208
SD	131	203	186	175	107	34	32

Table 13: Traverse distance of deformation (dL) at each 100 N for specimens of group LcT

Specimen Nr.	dL (100 N)	dL (200 N)	dL (300 N)	dL (400 N)	dL (100 - 200 N)	dL (200 - 300 N)	dL (300 - 400 N)
1	348	814	992	1183	466	177	191
2	553	770	940	1152	217	169	212
4	938	1354	1504	1710	415	151	206
5	595	870	1019	1201	275	149	181
7	490	764	883	1061	275	119	177
12	433	634	785	958	201	151	173
14	479	656	757	869	177	101	112
16	260	399	566	760	139	167	194
3	165	345	553	748	180	208	195
6	190	292	508	717	102	217	209
8	332	468	634	825	137	165	191
9	499	661	834	1025	163	173	191
10	271	410	573	751	139	164	177
11	67	157	279	398	90	121	119
13	473	782	943	1110	309	161	167
15	441	584	746	933	143	163	187
Median	437	645	771	945	179	163	189
Mean	408	623	782	963	214	160	180
SD	206	286	280	292	108	30	28