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Propädeutik und Werkstoffkunde
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Influence of zirconia ceramic abutment preparation on the fracture strength of single implant restorations after chewing simulation.

[Einfluss der Zirkonabutmentpräparation auf die Bruchfestigkeit von Einzelimplantatrestaurationen nach Kausimulation]

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List of Abbreviations

CAD/CAM : Computer Assisted Design/Computer Assisted Manufacturing

F: Cubic phase

LTD: Low Temperature Degradation

M: Monoclinic phase

Mg-PSZ: Magnesium- Partially Stabilized Zirconia

mm: millimeter

N: Newton

PSZ: Partially Stabilized Zirconia

T: Tetragonal phase

Y-TZP: Yttrium-Tetragonal Zirconia Polycrystals

Zr: Zirconium

1. INTRODUCTION

The concept of oral implants is based upon the process of osseointegration (Brånemark 1983). Resistance to bending forces in implants is essential for the long-term clinical performance of the restoration (McGlumphy et al. 1992, Basten et al. 1996). Clinical success depends upon many parameters such as connection design between implant and the abutment and the different materials used for the fabrication of either abutment or crown (Morgan et al. 1993, Tripodakis et al. 1995).

Oral implants in combination with titanium abutments and porcelain fused to metal crowns are the "gold standard" treatment option in prosthodontics with excellent survival rates (Lindh et al. 1998, Priest 1999, Zitzmann et al. 2001, Leonhardt et al. 2002). The technological evolution and development of innovative materials and techniques in implant dentistry has created a first class challenge for the clinical practitioner to evolve new clinical and esthetic standards. As a result, patients consider natural-looking teeth as an essential necessity of life and their tendency has focused on metal-free restorations (Meyenberg 1994, Vallitu et al. 1995). Therefore, optimal pink and white esthetics is a thought-provoking goal for both the surgeon and the restorative dentist (Zarb and Lewis 1992, Studer 1994). However, the color of the attached mucosa can be influenced by the implant and the abutment material especially in the case of thin periimplant tissues (Hürzeler et al. 1994, Studer et al. 1995). Abutment components often shine through the mucosa giving a grayish shade due the insufficient thickness of the periimplant soft tissues or the inadequate depth of the emergence profile (Prestipino and Ingber 1996). In addition, when metal abutments are used with all-ceramic crowns, the underlying metal receives, through the all-ceramic crown, a certain percentage of incident light altering the final color establishment of the all-ceramic restoration (Zarb and Lewis 1992, Hegenbarth 1997).

1.1. Ceramics

Ceramics are defined as non-metallic inorganic man made solid objects, formed by baking raw materials (minerals) at high temperature (Cronin and Cagna 1997, Rosenblum and Schulman 1997, Neumann 1999). According to the chemical composition dental ceramics can be classified into 3 categories: a) silicate ceramics (i.e. IPS Empress, Ivoclar Vivadent, Schaan, Liechtenstein; ProCAD, Ivoclar Vivadent; Vita Mark II, Bad Säckingen, Germany) b) non-silicate or high strength oxide ceramics (i.e. In-Ceram Alumina and Spinell (Vita, Bad Säckingen, Germany) Procera All Ceram (NobelBiocare, Goteborg, Sweden) and zirconium oxide ceramics and c) non-oxide ceramics such as nitrides and carbides (Blatz et al. 2003a, Neumann 1999). Based on the fabrication technique, all-ceramic restorations can be classified into five systems: traditional powder-slurry (i.e. Optec, Jeneric/Pentron Inc., Wallingford, CT, US.), castable ceramics (i.e. Dicor, Dentsply, York, PA, US), pressable ceramics (i.e. IPS Empress, Ivoclar Vivadent); machinable ceramics (i.e. Vita Blocks Mark II, Vita), and glass-infiltrated oxide ceramics (i.e. In-Ceram, Vita) (Cronic and Cagna 1997, Kelly et al. 1996, Rosenblum and Schulman 1997).

1.1.1. Pressable ceramics

Pressed ceramics such as the leucite reinforced glass-ceramic (i.e. IPS Empress, Ivoclar Vivadent) and lithium disilicate glass-ceramic (i.e. IPS e.max Press, Ivoclar Vivadent) use the principle of crystal dispersion. According to this principle a prefabricated colored ceramic ingot is heated and then pressed into an investment mold using a special furnace at a specific temperature (Evans and O'Brien 1999, van Djiken 1999). Dimensional changes occur during solidification of the molten glass-ceramic can be compensated by accurately matched expansion of the investment material (Kelly et al. 1996). Frameworks can be pressed either in a full anatomical contour which is completed by color staining or to a partial anatomical contour that supports porcelain veneering (Evans and O'Brien 1999). Leucite reinforced glass-ceramics may be used for laminate veneers, inlays, onlays, partial-coverage and complete crowns (Goulet 1997, Kelly 2004), while lithium disilicate glass-ceramics could be used for the same restorations plus three-unit anterior fixed dental prostheses (Djiken 1999, Kheradmandan et al. 2001). Due to the inherent glassy matrix of the intaglio glass-ceramic surface, it can be etched with hydrofluoric acid, silanated and bonded to tooth structure using resin cements (Özcan and Vallittu 2003, Filho et al. 2004).

1.2. Zirconium oxide ceramics (zirconia)

1.2.1. Medical and dental applications of zirconia

Zirconia-based materials were initially introduced for biomedical use in orthopedics for total hip replacement, because of their excellent mechanical properties and biocompatibility (Piconi et al. 1998, Piconi and Maccauro 1999). Zirconia ceramics have also been used in dentistry (Vagkopoulou et al. 2009, Koutayas et al. 2009) as crown and full (Fritzsche 2003, Burke et al. 2006) and partial (Komine and Tomic 2005, Wolfart et al. 2007) bridge framework material, as prefabricated post or/and core (Meyenberg et al. 1995, Koutayas and Kern 1999), as implant abutments (Wohlwend et al. 1996, Glauser et al. 2004) and as implants (Kohal and Klaus 2004). In addition, different zirconia dental auxiliary components where proposed for dental use as orthodontic brackets (Keith et al. 1994), precision attachments, cutting and surgical instruments.

1.2.2. General data of zirconia

Zirconium (Zr) is a metal (atomic number: 40) which was first discovered by the German chemist Martin Klaproth in 1789. It has a density of 6.49g/cm^3, melting point of 1,852°C and a boiling point of 3,580°C. It has a hexagonal crystal structure and a grayish color. Zirconium does not occur in nature in a pure state but it can be found as mineral in conjunction with silicate oxide (i.e. $ZrSiO_4$ known as Zircon) or as a free oxide (i.e. ZrO_2 known as Baddeleyite) (Lindemann 2000, Piconi and Maccauro 1999). Minerals cannot be used as primary materials in dentistry because of impurities of different metal elements that color their mass and natural radionuclides like urania and thoria, which make them radioactive. Therefore, complex and time-consuming purification processes are generated to produce pure zirconia powders that can be used for biomedical applications (Piconi and Maccauro 1999).

1.2.3. Stabilized zirconia

By the addition of stabilizing oxides to pure zirconia, such as calcium (CaO), magnesium (MgO), cerium (CeO_2) or yttria (Y_2O_3), material's phase transformations can be inhibited and therefore allow the production of a multiphase material, termed as stabilized zirconia, at room temperature (Christel et al. 1989, Piconi and Maccauro 1999).

Fully stabilized zirconia is produced, when more than 16mol% CaO (7.9wt %), 16mol% MgO (5.86wt %), or 8mol% Y_2O_3 (13.75wt %) is added into ZrO_2 and has a cubic form. However, the most useful mechanical properties can be obtained when zirconia is in a multiphase form, known as Partially Stabilized Zirconia (PSZ) (Garvie et al. 1975). Several PSZ have been tested as ceramic biomaterials. Mg-PSZ is one of the most commonly used zirconia-based engineering ceramics (Sundh and Sjogren

2006). It has been reported that reinforcement by phase transformation toughening is less pronounced in Mg-PSZ than in Y-TZP (see §.1.2.5. Yttrium-tetragonal zirconia polycrystals) (Sundh and Sjogren 2006). Ceria (Ce)-doped zirconia ceramics were rarely considered, although they exhibit superior toughness (up to 20 MPa√m) and show no aging (Chevalier 2006, Ban 2008).

1.2.4. Zirconia transformation-toughening mechanism

Zirconia can be found in three crystallographic phases: 1) the monoclinic phase (M) between room temperature and 1,170^{o}C, 2) the tetragonal phase (T) between 1,170 and 2,370^{o}C and 3) the cubic phase (F) above 2,370^{o}C (Figure 1.1).

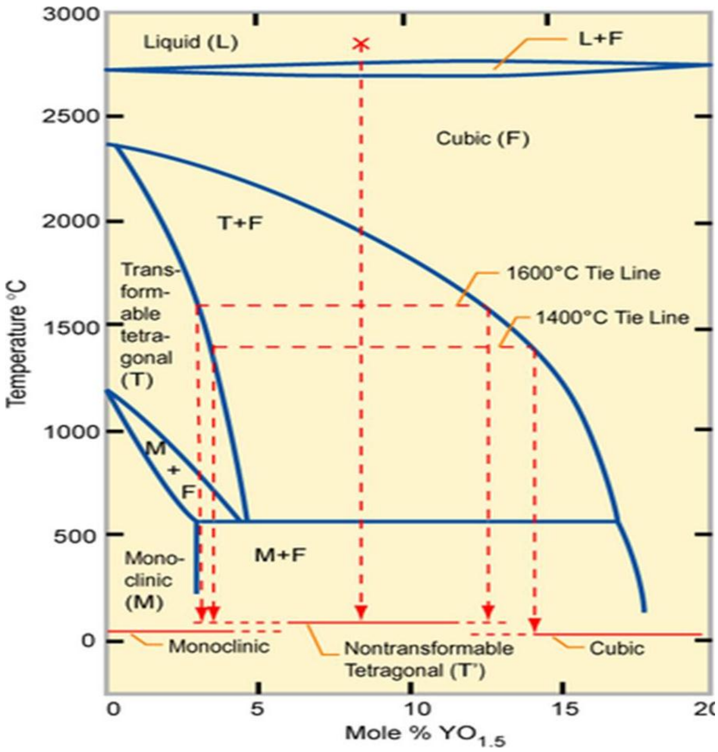

Figure 1.1. Phase relationship in zirconia-yttria systems according to composition and processing temperatures (M: monoclinic, T: tetragonal, F: cubic), *[Source: Piconi and Maccauro 1999]*.

In the presence of a small amount of stabilizing oxides such as Y_2O_3, it is possible to obtain at room temperature PSZ ceramics totally in the tetragonal phase, described as Tetragonal Zirconia Polycrystals (TZP). By the fine dispersion of stabilizing oxides grains within the cubic matrix, zirconia material can be maintained in a metastable state that able to transform into the monoclinic phase (Christel et al. 1989). This phenomenon can be explained through the lower surface energy of the tetragonal

Y_2O_3 particles and the constraint of the rigid matrix on them that opposes their transformation to the less dense monoclinic form. This process provides a powerful crack-inhibiting and therefore strengthening mechanism, termed as "*transformation toughening*" (Garvie et al. 1975). The tetragonal ZrO_2-grains can transform into the monoclinic phase when the constraint exerted by the matrix is relieved, i.e. during crack propagation (Piconi and Maccauro 1999). At the edge of the crack, a compressive stress field, associated with a 3 to 5% volume expansion of the transformed tetragonal grains, acts against the crack propagation (Figure 1.2) The fracture energy is dissipated both in the T-M transformation (also known as martensite-like transformation which occurs in quenched steel) and in the process of overcoming the compression stress of matrix due to the volume expansion. Therefore, the progression of the crack is inhibited and the toughness of the ceramic material is enhanced (Christel et al. 1989, Piconi and Maccauro 1999).

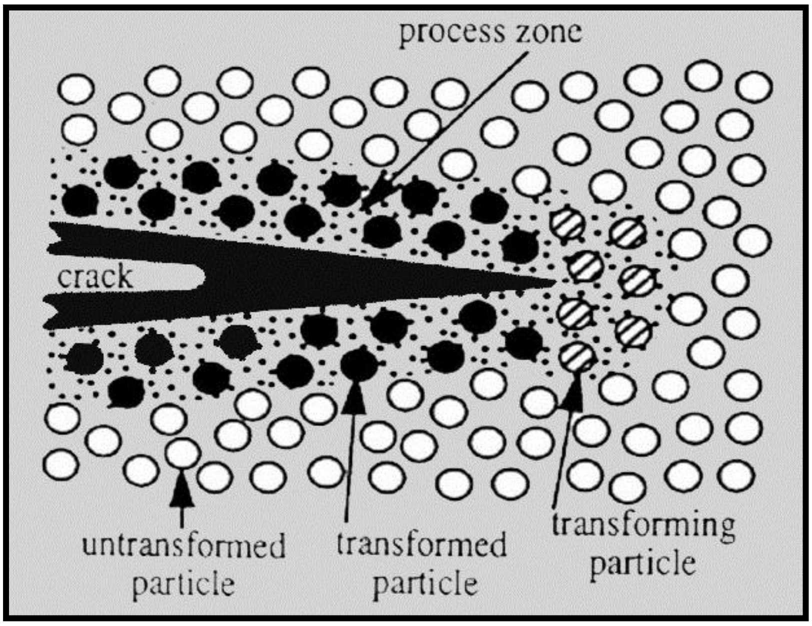

Figure 1.2. Schematic drawing of the stress-induced transformation-toughening mechanism in TZP [Source: Piconi and Maccauro 1999].

1.2.5. Yttrium-tetragonal zirconia polycrystals (Y-TZP)

The addition of approximately 2-3% mol yttria (Y_2O_3) as a stabilizing agent in zirconia allows the sintering of fully tetragonal zirconia ceramic material, made of 100% metastable tetragonal grains, known as *"yttrium-tetragonal zirconia polycrystals"* (Y-TZP) (Christel et al. 1989). The amount of the T-phase at room temperature and therefore the mechanical properties of Y-TZP ceramics, are relative to the yttrium content and grain size, the processing temperature and finally the constraint exerted on them by the matrix (Piconi and Maccauro 1999). The addition of Y_2O_3 in higher concentrations produces a fully stabilized zirconia ceramic in a complete cubic phase which presents lower fracture characteristics (Sato and Shimada 1985b). To obtain a metastable tetragonal structure at room temperature such as 3mol% Y_2O_3-ZrO_2, the ceramic grain size must be less than 0.8 µm (Theunissen et al. 1992). A critical grain size exists, linked to the yttria concentration, above which spontaneous T-M transformation of grains takes place whereas this transformation would be inhibited in an overly fine-grained structure (Picconi and Maccauro 1999) (Figure 1.3).

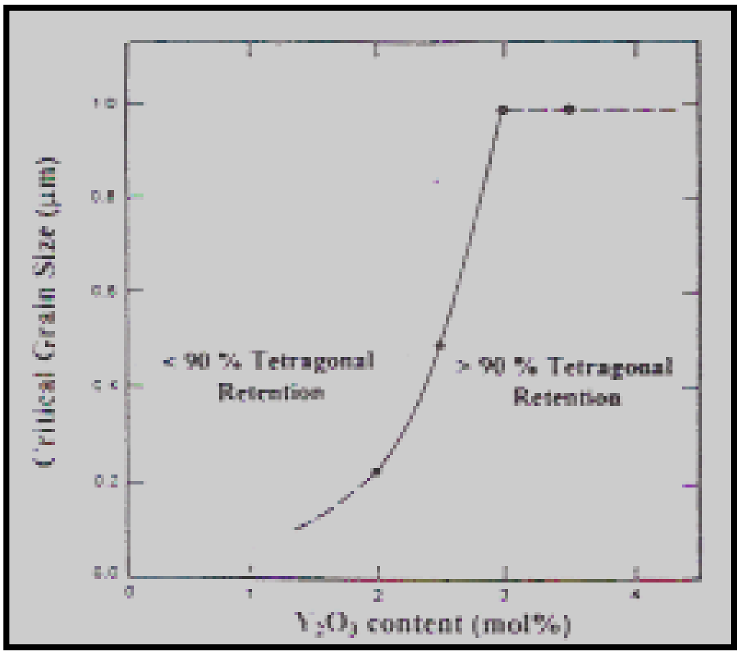

Figure 1.3. Retention of tetragonal phase. Critical grain size against yttria content in TZP *[Source: Piconi and Maccauro 1999]*.

Finally, the T-M transition in Y-TZP materials does not only depend on the Y_2O_3 content, but also on its distribution in the material's mass. For his reason, the stabilizing oxide should be added during the early stages of the ceramic powder manufacturing process. Alternatively, it can be co-precipitated with ZrO_2 salts or coat the ZrO_2-grains during zirconia powder production (Piconi and Maccauro 1999).

1.2.6. Chemical and physical properties of Y-TZP

The chemical and physical properties of Y-TZP are listed below in Table 1.1.

Table 1.1. Chemical and physical properties of Y-TZP [Source: Vakgokpoulou et al. 2009].

Properties	Y-TZP
Chemical composition (wt %)	
$ZrO_2 + HfO_2 + Y_2O_3$	>99.0
Y_2O_3	4.5 to 5.4
Al_2O_3	<0.5
Other oxides	<0.5
Physical properties	
Density (g/cm^3)	6.05
Grain size (µm)	0.2
Monoclinic phase (%)	1
Porosity	<0.1 %
Mechanical properties	
Flexural strength [4 point] (MPa)	1,666.0
Elastic modulus (GPa)	201
Vickers hardness (HV)	1,270.0
Fracture toughness (Kgf/mm$^{2/3}$)	16.8
Fracture toughness KIC (MPa m^{-1})	7-10
Compressive strength (MPa)	4,900.0
Impact strength (MPa)	137.0
Thermal properties	
Thermal expansion coefficient (x10^{-6}/°C)	11x10^{-6} K^{-1}
Thermal conductivity (W/m°K)	2
Thermal Shock Resistance (ΔT °C)	280-360
Specific heat J/kg°K	500
Electrical properties	
Dielectric constant (1MHz @ R.T.)	26 @100kHz
Dielectric Strength (kV/mm)	9.0
Electrical Resistivity (Ωcm @ R.T.)	>10^{13}
Optical properties	
Refractive index	2-2.2
Light transmittance	<48%

1.2.7. Biocompatibility of Y-TZP

A small percentage of the population is hypersensitive to dental alloys (Hansen and West 1997). In vitro and in vivo studies have confirmed the high biocompatibility of Y-TZP when high purity zirconia-powders are used. As a result, no local (cellular) or systemic adverse reactions have been reported (Christel et al. 1989, Covacci et al. 1999, Josset et al. 1999, Piconi and Maccauro 1999). Recent studies have demonstrated that lower plaque accumulation around Y-TZP than titanium restorations (Rimondini et al. 2002, Scarano et al. 2004). This has led to the suggestion that zirconium oxide may be a suitable material for manufacturing implant abutments that exhibit reduced bacteria colonization potential (Scarano et al. 2004).

1.2.8. Aging of Y-TZP

Long-term stability of ceramics depends on the subcritical crack growth and the stress corrosion caused by water. ZrO_2-ceramics are prone to age in wet environment, which is of particular concern for biomedical applications, because results to degradation of their mechanical properties (Chevalier 2006). This specific aging phenomenon is termed as *"low temperature degradation (LTD)"* of zirconia occurs because of a progressive spontaneous transformation of the tetragonal phase into monoclinic which leads to surface damage when Y-TZP is in contact with water or vapor (Sato and Shimada 1985b), body fluid, or during steam sterilization (Piconi and Maccauro 1999). In addition, non-aqueous solvents with a chemical structure similar to water can also destabilize Y-TZP, causing strength degradation (Sato and Shimada 1985b, Ardlin 2002).

According to Swab (1991), the critical points of Y-TZP aging are the following: 1) The most critical temperature range for this phenomenon is 200-300°C, 2) The effects of aging are the reduction in strength, toughness and density and an increase in monoclinic phase content. 3) The degradation of the mechanical properties is due to the T-M transition and is taking place with micro and macro cracking of the material, 4) T-M transition starts on the surface and progresses inwards the mass of the material, 5) grain size reduction and/or concentration increase of the stabilizing oxide reduce the transformation rate, and 6) T-M transformation is enhanced in water or in vapor.

The formation of Zr-OH bonds accelerates crack growth of pre-existing flaws and promotes the T-M phase (Sato and Shimada 1985b, Piconi and Maccauro 1999).

LTD results in surface degradation of zirconia, in terms of: a) roughening, which leads to increased wear, and b) microcracking, which leads to grain pullout, generation of particle debris and possible premature failure. Surface elevations take place because of the voluminous M-phase and depend on the different aging medium (Ardlin 2002). Apart from elevations, craters have also been observed, as a result of worn out monoclinic spots on the degraded surface of the material (Chevalier 2006).

LTD rate of Y-TZP is related to several factors, such as chemical composition, duration of exposure to aging medium, loading of the ceramic restoration and manufacturing processes, all of which affect the microstructure of the material (Ardlin 2002, Chevalier 2006).

In regard to Y-TZP chemical composition, changes in yttrium concentration and processing temperature define the amount of the tetragonal phase in the material and, thus, the amount of the transformed M-phase. It has been suggested that the initial amount of monoclinic phase should be less than 10% for every surface of the material in contact with body fluids (Chevalier 2006). In addition, reduction of grain size reduces the transformation rate (Sato and Shimada 1985b). Dramatic decrease of the LTD resistance has been observed, when the grain size was greater than 0.6μm. However, after the initial phase-transformation, a stable state with no further significant decrease in flexural strength (>700MPa) can be established (Ardlin 2002, Shimizu et al. 1993).

1.2.9. Y-TZP surface and heat treatments

Processing and veneering of zirconia-based frameworks, fabricated with CAD/CAM technology, involves different laboratory stages such as grinding, polishing, air-borne particle abrasion and heat treatment. The critical influence of these stages on the aging sensitivity of zirconia affects the long-term stability and success of the material.

Grinding. Generally grinding increases the strength of ceramics that contain metastable tetragonal zirconia. This is due to the T-M transformation on the surface of the material and the development of compressive strains from the transformation-related volume increase, at a depth of several microns under the surface (Garvie et al. 1975). The surface compressive stresses prevent microcrack formation or propagation, but promote surface and subsurface damage. After zirconia phase transformation, surface damage mechanism is by grain pullout and formation of microcraters (Denry and Holloway 2006). This leads to surface roughness and porosities which may influence the wear resistance of the material (Piconi and Maccauro 1999). Apart from the strained tetragonal grains, a rhombohedral zirconia phase has been found to form after grinding, with similar consequences on the behavior of zirconia as the tetragonal phase (Denry and Holloway 2006).

Despite water-cooling, high stresses and temperatures are developed during grinding. These high temperatures are especially generated during machine grinding than during manual grinding (Swain and Hannink 1989), and can activate the reverse M-T transformation (Ardlin 2002, Kosmac et al. 1999). For this reason, a certain amount of M-phase is removed from the surface and, thus, material's strength is reduced. The use of water spray during grinding reduces stresses, resulting in a decrease of the critical flaw size by about 30% (Kosmac et al. 1999). Manual grinding is performed with less stress and at lower temperatures; therefore, it promotes the T-M transformation and increases the surface compressive layer (Ardlin 2002).

Annealing after grinding may reverse zirconia M-T transformation but surface and subsurface damage remains and could subsequently lead to crack propagation (Denry and Holloway 2006). Furthermore, the introduction of deep surface flaws during machining (CAM) may concentrate stresses that also determine strength of the restoration (Kosmac et al. 1999, Luthardt et al. 2004). Grinding of the inner surface of zirconia crowns induces surface flaws and microcracks at the internal surface of the occlusal region. As shown in failed restorations, these areas concentrate the greatest tensile stresses during clinical loading. Thus, it is important that the concentration of microcracks in these areas is minimized (Luthardt et al. 2004). Moreover, coarse grinding tools that produce deep surface flaws and extensive heat may also determine the strength of the restoration (Ardlin 2002, Kosmac et al. 1999, Luthardt et al. 2004). The direction of tool rotation during machining and the sharpness and number of active diamond grains seem to be important determinants of surface properties of the material (Luthardt et al. 2004). Orientation of grinding was found irrelevant; however, fractures may be initiated under loading by the flaws which distributed perpendicular to the grinding orientation (Guazzato et al. 2005).

Polishing. The polishing process develops scratches that induce residual stresses in the material. The influence of polishing on the aging sensitivity of zirconia is contradictory and relates to the type and amount of these stresses.

Rough polishing produces a compressive surface stress layer beneficial for the aging resistance, while smooth polishing produces preferential transformation nucleation around scratches, due to tensile residual stresses caused by elastic/plastic damage (Deville et al. 2006). Fine polishing after grinding may remove the compressive layer of monoclinic phase from the surface, while further polishing may minimize the size of flaws and result in greater flexural strength (Guazzato et al. 2005).

Air-borne particle abrasion. Air-borne particle abrasion of the inner surface of a restoration is usually used to enhance the adhesion strength of the luting agent to the framework (Kern and Wegner 1998). According to Kosmac et al. (1999), it can also provide a powerful technique for strengthening Y-TZP at the expense of somewhat lower stability. The alumina particles can cause significant damage to the material's surface, which is characterized by erosive wear and lateral cracks. However, a thin compressive layer of transformed M-phase is formed, which counteracts the strength degradation caused by air-borne particle abrasion-induced flaws and effectively increases strength. Lower temperatures and stresses are developed than in grinding allowing the M-phase to be maintained (Guazzato et al. 2005).

The use of air-borne particle abrasion after grinding may reduce the critical size of flaws through chipping, which largely levels the material surface (Kosmac et al. 1999). On the other hand, it has been found that air-borne particle abrasion before cementation of Y-TZP restorations mechanically assists the growth of pre-existing flaws, reducing the strength the material when compared to polished specimens (Zhang et al. 2004). Nevertheless in this research no restorations were tested and the inner surfaces of restorations are never polished but always machined.

Heat treatment. Surface and heat treatments have a counteracting effect on flexural strength of dental Y-TZP ceramics (Guazzato et al. 2005). While the strength of Y-TZP can be increased by wet grinding or air-borne particle abrasion, when followed by heat treatment it is reduced. The effect of heat, regardless of the holding time, initiates the reverse M-T transformation, eliminating the M-phase from the material surface and thus lowering the strength of the material (Guazzato et al. 2005).

Clinically, a greater amount of monoclinic phase on the surface and therefore a greater flexural strength may be desirable. On the other hand, an excessive amount of M-phase could lead to microcracking (Guazzato et al. 2005) and predispose the material to a more rapid moisture-assisted transformation over time and loading in the acidic and aqueous oral environment (De Aza et al. 2002).

1.3. Ceramic abutments

In the 90's, individualized ceramometal abutments offered an esthetic approach for single implant crowns (Prestipino and Ingber 1993a, Pröbster and Groten 1997, Marinello and Meyenberg 1997). However, with the introduction of high toughness ceramics different all-ceramic abutments with improved physical and optical properties became available for dental use (Prestipino and Ingber 1993b). The bio-esthetic outcome of all-ceramic abutments in combination with all-ceramic crowns has been demonstrated in several clinical trials (Studer and Wohlwend, Wohlwend and Studer 1996, Sadoun and Perelmuter 1997, Heydecke et al. 2002, Bonnard et al. 2001). All-ceramic abutments could positively contribute to the final color establishment of an all-ceramic restoration. This specific potential is related to a deeper diffusion and absorption of the transmitted light into the ceramic abutment mass, which approximates the translucency of the natural teeth (Pietrobon and Paul 1997, Koutayas and Kern 1999, Tan and Dunne 2004).

Current ceramic abutments are fabricated from either densely sintered high-purity alumina (Al_2O_3) ceramic (Dahlmo et al. 2001) or partially stabilized yttria (Y_2O_3)-tetragonal zirconia (ZrO_2) polycrystal (Y-TZP) ceramic (Glauser et al. 2004). Both materials demonstrate differences regarding their microstructure and mechanism against flaw propagation (Christel et al. 1989, Andersson and Oden 1993, Mante et al. 1993, Seghi et al. 1995). Y-TZP abutments presented 3-fold improved fracture strength than the alumina ones (Yildirim et al. 2003). The main disadvantage of the alumina abutments is related to reduced strength and fatigue resistance when compared to metal abutments (Ingber and Prestipino 1991a, Ingber and Prestipino 1991b, Prestipino and Ingber 1993).

1.3.1. Zirconia implant abutments

Expected high survival rates for implants and implant-supported single crowns (Jung et al. 2008) can be also accommodated by the clinical application of zirconia abutments (Kohal et al. 2008). Moreover, Y-TZP abutments may promote soft tissue integration (Welander et al. 2008) and provide clinically favourable peri-implant soft tissues (Glauser et al. 2004, Degidi et al. 2006, Bae et al. 2008). A systematic review revealed that Y-TZP abutments, compared to Ti or Au alloy and alumina ones, could equally preserve the peri-implant bone level (Linkevicius and Apse 2008).

Prefabricated Y-TZP abutments are commercially available from different implant manufactures or can be fabricated customized by dental technicians. Regarding the latter, CAD/CAM technology can be beneficial in designing fully individualized Y-TZP abutments. Selected prefabricated and custom-made Y-TZP abutments are shown in Table 1.2. Both types of abutments can be further customized either by extraoral or intraoral preparation.

Most manufactures recommend either a pronounced chamfer or a shoulder preparation with rounded inner line angles. Subgingival preparation margins should not be overextended beyond the point that provides access for cleaning (i.e. cement residuals). Moreover, the emergence profile especially for customized abutments should be rather concave (Rompen et al. 2007) and must follow known diagnostic regimens (Yildirim et al. 2000). Quasi-static loading testing (loading direction of 60º) after 0.5 and 1 mm of axial reduction of zirconia abutments (AstraTec) did not significantly affect the fracture resistance (between 429 and 576 N) of single implant crowns which was gathered above the estimated anterior incisive forces (Adatia et al. 2009).

Marginal adaptation of zirconia abutments can be achieved either by the abutment with or without an integrated titanium basis and a fastening screw (Brodbeck et al. 2003) In-vitro precision fit evaluation of internal or external hexagon CAD/CAM custom abutments met clinical standards (Lang et al. 2003) and in case of hexagonal external connection showed less than 3º of rotational freedom (Vigolo et al. 2006). Screw joint ceramic abutments may present fracture or loosening implications due to misfit at the implant/abutment interface (Tripodakis et al. 1995, Papavasiliou et al.1996) should be avoided through appropriate laboratory processing (Vigolo et al. 2005).

In-vitro testing of CAD/CAM-processed implant single crowns supported by either prefabricated (Butz et al. 2005, Att et al. 2006a, Gehrke et al. 2006, Att et al. 2006b) or customized (Sundh and Sjogren 2006) Y-TZP abutments confirmed their feasibility to withstand physiologic incisive forces. Additional dynamic loading results using a chewing simulator, regarding of implant single crowns on Y-TZP abutments (Butz et al. 2005, Att et al. 2006a, Att et al.2006b, Gehrke et al. 2006) were also confirmed by clinical research that revealed cumulative survival rate of 100% up to 6 years of service. However due to the lack of the number and the moderate observation time of the existing clinical studies, further long-term evaluation is necessary (Linkevicius and Apse 2008).

Table 1.2. Selected zirconia abutments [Source: Koutayas et al. 2009].

Manufacturer	Friadent		Nobel Biocare		Straumann	Biomet 3i		Bego	Astra	Atlantis
www.	friadent.de		nobelbiocare.com		straumann.com	biomet3i.com		bego.com	astratechdental.com	atlantiscomp.com
Name	Cercon Balance	Friadent Cercon	Procera Abutment Zirconia	Procera Abut. Zirconia for other implants	RN SynOcta custom abut. [CARES]	Ext. hex. ZiReal post	Certain ZiReal post	BeCe Sub-Tec ceramic	Zirdesign	Atlantis
Material	Y-TZP	Y-TZP	Y-TZP	Y-TZP Ti seating post	Y-TZP Ti seating post	Y-TZ P Ti seating ring or post		Y-TZP	Y-TZP	Y-TZP
Color	whitish	whitish, dentin	whitish	Whitish	Whitish	whitish		Whitish	whitish	whitish
Connection	int. cone Ti screw	int. hex Ti screw	exτ. hex Ti screw	int. hex Ti screw	int. hex (SynOcta 1.5) Ti screw	exτ. hex Au-screw	int. hex Au-screw	int. cone & hex & Ti screw	int. cone & hex & Ti screw	int. hex or ext. hex & system screw
Implant Diameter (mm)	Ankylos 5.5, 7.0	XiVe 3.8, 4.5	all Brånemark NP / RP / WP	NobelReplace NP / RP / WP Straumann RN 4.8, Camlog 3.3- 6.0	Straumann RN 4.8	Osseotite nt, pw, xp 4.1, 5.0	Osseotite certain 4.1, 5.0	Bego S 3.25-5.5 Begor R 3.75-5.5	Osseospeed 3.5/4.0, 4.5/ 5.0	int. hexed impl. ext. hexed impl.
Gingival Height (mm)	1.5, 3.0 scalloped	1.0, 2.0	Customized		Customized	4.0			1.5, 3.0 scalloped	customized
Inclination	straight (0º), angled (15º)	straight (0º), angled (15º)	Customized		Customized	straight (0º)		straight (0º)	straight (0º), angled (20º)	customized
Type	prefabricated	prefabricated	customized (Procera 3-D CAD)		customized Sirona InLab	prefabricated		Prefabricated	prefabricated	customized (Atlantis VAD)

2. PURPOSE

Zirconia ceramic seems to be a very promising biomaterial for the fabrication of implant abutments. However, scarce data can be found in the literature regarding the influence of the preparation depth and design on the biomechanical and the feasibility behaviour of all-ceramic crowns luted to Y-TZP abutments. In addition, clinical evaluation in the highly loaded oral environment requires long-term studies which are costly and time-consuming.

The hypothesis to be tested was that the increase of the preparation depth and the manual, instead the manufacturer milling, preparation mode of zirconia implant abutments will not negatively affect the fracture strength of lithium disilicate glass ceramic implant crown restorations under different loading conditions (quasi-static and dynamic loading).

The purpose of the study was to evaluate the influence of the circumferential chamfer preparation depth (0.5, 0.7, 0.9 mm) using two preparation modes (milling by the manufacturer, milling by Celay system) on the fracture strength, and to explore the fracture mode of lithium disilicate glass ceramic crowns (IPS e.max Press, Ivoclar Vivadent) on single implant zirconia abutments (ZirDesign, Astra Tech, Mölndal, Sweden) under different loading modes (dynamic, quasi-static loading).

3. MATERIALS AND METHODS

3.1. Materials

All materials used for the study are listed in Table 3.1.

Table 3.1. Study materials.

Material	Manufacturer	Generic Name	LOT No.
IPS e.max Press	Ivoclar Vivadent, Schaan, Liechtenstein	Lithium disilicate glass-ceramic	H21370
OsseoSpeed	Astra Tech, Mölndal, Sweden	Titanium alloy implant	45334
ZirDesign Type A, Ø 3,6 / Ø 4,1 Type B, Ø 3,2 / Ø 3,7	Astra Tech, Mölndal, Sweden	Y-TZP implant zirconia abutments	46952 Charge No. 6417 Index YA06 Charge No. 6417 Index YA06
ZirDesign Abutment screw	Astra Tech, Mölndal, Sweden	Titanium Screw	53758
Multilink Automix	Ivoclar Vivadent, Schaan, Liechtenstein	Dual curing resin cement	J05820
Metal Zirconia Primer	Ivoclar Vivadent, Schaan, Liechtenstein	Coupling Reagent	H36277
Steatite Ceramic Ball	Höchst Ceram Tec, Wunsiedel, Germany	Steatite Ceramic	-
Technovit 4000 Powder Technovit Syrup I Technovit Syrup II	Heraeus Kulzer, Wehrheim, Germany	Self-curing polyester resin	010221 011020 012010
Zwick Z010/TN2	Zwick, Ulm, Germany	Universal testing machine	-
Celay System	Mikrona, Spreitenbach, Switzerland	Copy-milling Machine	-
Celay Milling pins YZ-54S	Mikrona, Spreitenbach, Switzerland	Diamond cutting instruments	E284021
Willytec Chewing Simulator	Willytec, Munich, Germany	Chewing Simulator	-
Thermocycling apparatus	Gebrüder Haake GmbH, Karlsruhe, Germany	Thermocycling Apparatus	-

3.1.1. ZirDesign (Astra Tech, Mölndal, Sweden)

ZirDesign (Astra Tech) is an yttria tetragonal zirconia polycrystal material which can be further customized modified through essential preparation to the desired anatomical demands. This specific transmucosal abutment is used for cement-retained restorations (i.e. all-ceramic crowns) in combination to a corresponding implant. This ivory colored ceramic abutment is used for implant supported restorations placed in the anterior, canine and first premolar regions and strives to serve high esthetic demands. According to the manufacturer, it has a bending strength between 1,000 and 1,300 MPa, a fracture toughness of 9 to 10 MPa m⅓, a modulus of elasticity of 210 GPa and a linear thermal expansion coefficient of 10.6 x 10^{-6} K^{-1}. Detailed dimensions of this zirconia abutment and screw are shown in Figure 3.1.

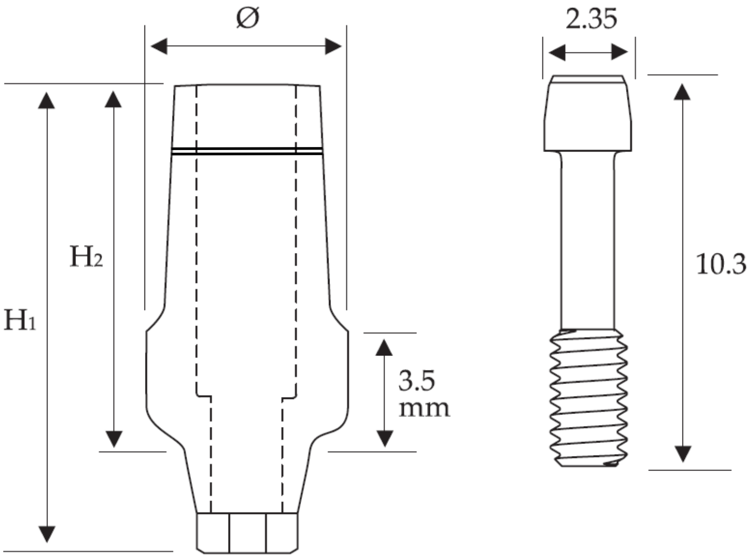

Fig. 3.1. Schematic drawing and dimensions of the abutment and screw. [H1:13.7 mm, H2= 10 mm, gingival height: 3.5 mm. screw diameter: 2.35 mm, screw length: 10.30 mm].

3.1.2. IPS e.max Press (Ivoclar Vivadent, Schaan, Liechtenstein)

IPS e.max Press (Ivoclar Vivadent) is a lithium disilicate glass-ceramic that is manufactured in ingots of two different sizes and four opacity levels (LT, MO, HL, HO) each. The material contains different chemical compounds such as SiO_2, Li_20, K_2O, P_2O_5, ZrO_2, ZnO, Al_2O_3, MgO, La_2O_3 and pigments. The ingots exhibit an optimized homogeneity, which results in a consistently high strength of 400 MPa. Moreover, IPS e.max Press follows the well established heat pressing technique. The completed core generally provides a desirable depth of translucency that facilitates an esthetic outcome after veneering. IPS e.max Press crowns are recommended as single tooth restorations for all intraoral regions.

3.1.3. Multilink Automix (Ivoclar Vivadent, Schaan, Liechtenstein)

Multilink Automix (Ivoclar Vivadent) is dual curing resin cement (self-curing luting composite with light-curing option) mainly composed by hydrolytically stable phosphonic acids (acidic monomers). The monomer matrix is composed of 22 to 26% dimethacrylate, 6-7% HEMA and 1% is benzoylperoxide. The inorganic fillers (40% in volume) are barium glass, ytterbium trifluoride and spheroidmixed oxide. Specifically, this cement has a particle size of 0.25-3.0 microns (mean particle size 0.9 microns). Multilink Automix 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, bridges and endodontic posts.

3.1.4. OsseoSpeed (Astra Tech Dental, Mölndal, Sweden)

OsseoSpeed implant (Astra Tech) is a two-piece system suitable for both one-stage and two-stage surgery. It has an improved titanium dioxide blasted surface which is fluoride-modified that as claimed by the manufacturer can rapidly stimulate bone formation. Furthermore, it has the ability to provide an increased bone-to-implant contact ratio and a stronger bone-to-implant interface. In addition, the neck is designed with MicroThread™ minute threads that offer optimal load distribution and lower stress values. Implant abutment is fixed into the implant with a conical connection (Conical Seal Design™), which is a below the marginal bone level and therefore transfers functional loads deeper down in the bone. Last but not least, after abutment connection a special contour is achieved that allows for an increased connective soft tissue contact zone both in height and volume, which integrates with the transmucosal part of the implant, sealing off and protecting the marginal bone. OsseoSpeed implants (Astra Tech) are commercially available in four (4) diameters (3.0, 3.5, 4.0, 4.5, 5.0 mm) and eight (8) different lengths (6.0, 8.0, 9.0, 11.0, 13.0, 15.0, 17.0, 19.0 mm). More specifically, OsseoSpeed implants with a 4.5 mm in diameter and 15.0 mm in length were used in this study.

3.2. Methods

3.2.1. Study outline

Seventy single implant-supported all-ceramic crowns were manufactured using a lithium disilicate glass ceramic material (IPS e.max Press, Ivoclar Vivadent) to replace a maxillary central incisor. For the purposes of the study, a full wax-up of a right maxillary central incisor (11) was fabricated (11.0 mm in height and 8.0 mm in width). For the study purposes, 56 zirconia abutments (ZirDesign, Astra Tech) were prepared in two depths (0.7 mm, 0.9 mm) following two preparation modes (milling by the manufacturer, milling by the Celay system). Additional abutments (n=14) that had been prepared by the manufacturer in the depth of 0.5 mm served as control. After the zirconia abutments were connected to 70 identical implants (Osseospeed, Astra Tech), 4.5 mm in diameter and 15.0 mm in length, all crowns were adhesively cemented (Multilink Automix, Ivoclar Vivadent). Study specimens were divided into five (5) groups of 14 specimens each. Subgroups of 7 specimens each were finally subjected to either quasi-static or dynamic loading test under 135° (Table 3.2.). Fracture strength values and number of loading cycles were recorded and statistically evaluated. Fracture modes were evaluated under low power stereo-magnification using an optical microscope.

Table 3.2. Test groups.

Group	Preparation Depth (mm)	Preparation Mode (milling)	Quasi-static Loading (S)	N	Dynamic Loading (C)	n
A	**0.5**	**manufacturer**	**SA**	**7**	**CA**	**7**
B	**0.7**	**manufacturer**	**SB**	**7**	**CB**	**7**
C	**0.9**	**manufacturer**	**SC**	**7**	**CC**	**7**
PB	**0.7**	**Celay system**	**SPB**	**7**	**CPB**	**7**
PC	**0.9**	**Celay system**	**SPC**	**7**	**CPC**	**7**
Partial Sum				**35**		**35**
Total Sum						**70**

3.2.2. Abutment preparation

The concept of this study was based on simulating and testing single implant crowns after different abutment preparation depths that can be clinically selected. The specific abutments are delivered by the manufacturer with a 0.5 mm circumferential chamfer margin. However; clinically this may be not enough and therefore can be further prepared both intraorally and extraorally by either the clinician or the technician, respectively. The selected circumferential chamfer preparations were extended 0.2 and 0.4 mm in depth from the original abutment size. Regarding the maximum radius of the abutment (1/2 of the diameter), the 70 abutments used in this study were finally prepared as follows: 1) Preparation depth (mm): 0,5 mm [Group A (n=14, control)], 2) Preparation depth (mm): 0,7 mm [Group B (n=28)] and 3) Preparation depth (mm): 0,9 mm [Group C (n=28).]. A schematic drawing of all different abutment preparations is shown in Figure 3.2.

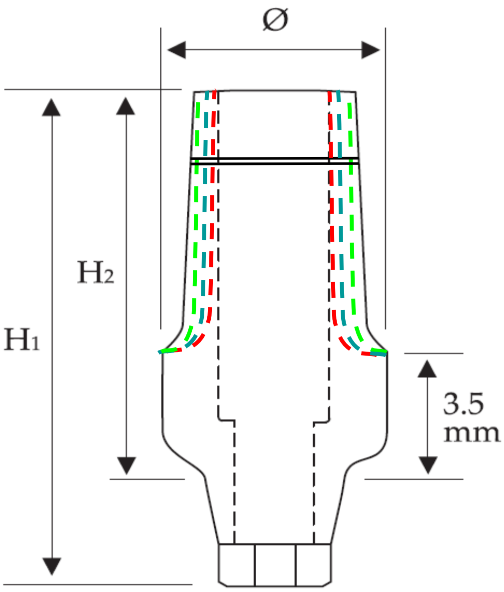

Figure 3.2. Schematic drawing of the three different prepared abutments.

: 0.5 mm reduction from original abutment size (Group A)

: 0.7 mm reduction from original abutment size (Group B)

: 0.9 mm reduction from original abutment size (Group C)

Appropriate abutment preparations were performed by milling either following the production line used for all commercially available abutments from the manufacturing company (Astra Tech) or using the Celay system (Mikrona) in a laboratory environment. Therefore, all abutments in Group A (n=14, control) and half of the abutments in Group B (n=14) and Group C (n=14) received the aforementioned axial preparations (milling) without height reduction by the manufacturer (Figure 3.2.).

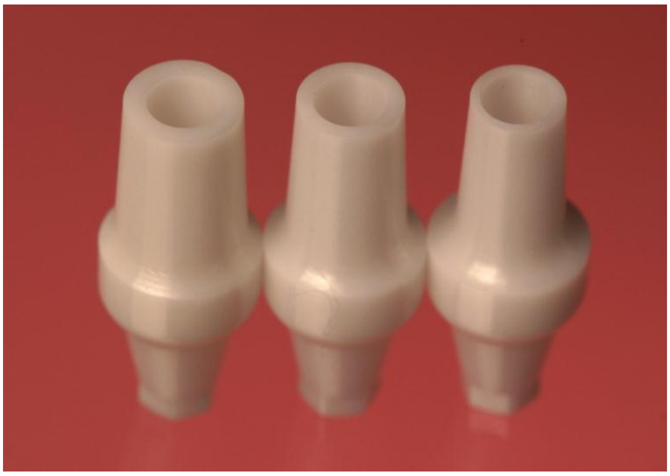

Figure 3.2.
Abutments prepared by the manufacturer (Astra Tech) without height reduction: Group A (left), Group B (middle), Group C (right).

Height modifications were made using the Celay system (Mikrona) (Figure 3.3.a.) using special cutting diamond instruments (Figure 3.3.b.) under water coolant. The incisal edge of the abutments was reduced in height so that each prepared abutment was 5.0 mm at the labial and 3.0 mm at the palatal aspect.

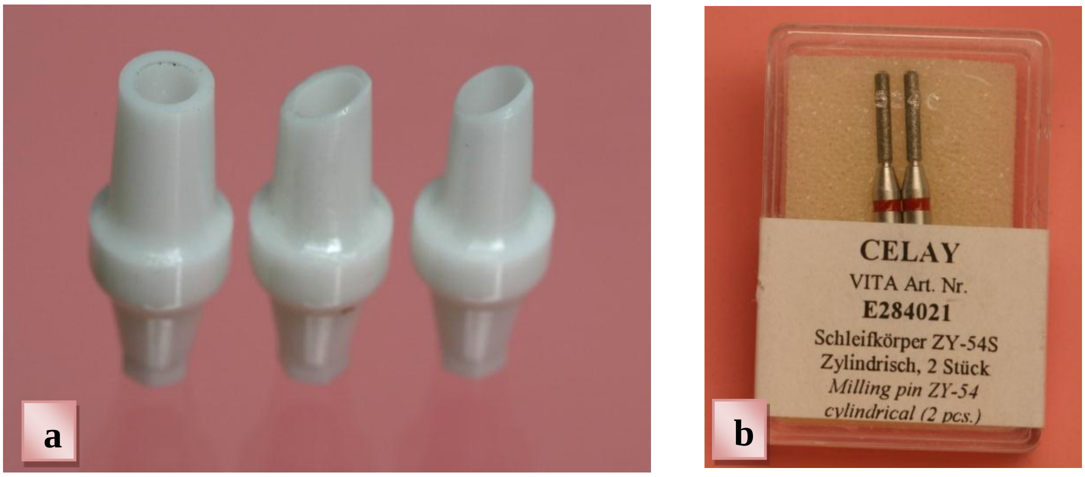

Figure 3.3.a.: Prepared abutments made by the manufacturer after modification using the Celay system: Group A (left), Group B (middle), Group C (right). Figure 3.3.b: Fine grained cutting diamond instruments especially designed for the Celay system.

Moreover, the additional half of the abutments in Group PB (n=14) and Group PC (n=14) received a manual preparation (milling) using the Celay system (Mikrona) and its special cutting diamond instruments under water coolant (Figure 3.3.b) and according the specification described previously (preparation depth of Group PB: 0.7 mm and of Group PC: 0.9 mm, abutment height: 6.0 mm).

Generally each zirconia implant abutment received a 360° circular chamfer preparation with rounded inner angles to the selected depth using the appropriate rotating instruments. All prepared abutments had a standardized 6° convergence and angle surfaces between the axial and palatal surfaces were rounded, as well as the incisal surfaces (minimum radius: 0.5 mm). However, a minimum width of 1.0 mm of the incisal edge in the vestibular-oral direction was retained to guarantee an exact reproduction of the internal framework surfaces by the milling unit.

In order to achieve identical dimensions during preparation of the abutments with the copy-milling technique, a pre-prepared to the selected preparation size and depth master metal abutment was attached on the tracing chamber of the Celay system (Mikrona). Finally, 4.5 mm milling implant analogues were use to facilitate tracing and milling purposes.

3.2.3. Fabrication of the master dies

Prior to the fabrication of the master dies all prepared zirconia abutments were connected onto identical titanium implants (Osseospeed, Astra Tech), 4.5 mm in diameter and 15.0 mm in length. According to the manufacturer's recommendation, every abutment was fixed with a standard abutment titanium screw (2.35 mm in diameter, 10.30 mm in length) using a torque control screw driver with a torque of 25 N/cm.

Then, the implant/abutment specimens were embedded in a three-component, self-curing polyester resin (Technovit 4000, Heraeus Kulzer, Wehrheim, Germany) using a preset silicon index that provided a horizontal inclination of 135°. Polyester resin material was poured into special cooper cylinders that also served as the specimen holders during testing.

3.2.4. Fabrication of the crowns

For the fabrication of the 70 lithium disilicate crowns, full wax-ups of the complete crown restorations were made onto the zirconia abutments in order to replace a right maxillary central incisor (11). Identical wax-ups were performed with respect to the external crown dimensions such as 11.0 mm in height and 8.0 mm in width. The latter were achieved with the use of a silicon index, which was taken from a master diagnostic wax-up, and verified, with the use of a digital caliper. After burn out of the wax crown analogue, a castable lithium disilicate glass-ceramic ingot (IPS e.max Press, Ivoclar Vivadent) was heated and pressed into an investment mold using the heat-pressing technique. Finally, all crowns were fitted down to the master dies and completed by appropriate grinding and polishing.

3.2.5. Bonding procedure

For the adhesive cementation, the zirconia implant abutments were air-abraded with 50 μm alumina particles at 0.5 bars pressure until a marker coating (green) was completely removed. Moreover, they were ultrasonically cleaned in alcohol 96% for 2 minutes, dried and pre-treated with a special primer (Metal-Zirconia primer, Ivoclar Vivadent). In addition, the inner surfaces of the lithium disilicate crowns were etched according to the manufacturer for 20 seconds with hydrofluoric acid (IPS Ceramic Etching Gel, Ivoclar Vivadent) and silanated (Monobond S, Ivoclar Vivadent). Then the crowns were bonded onto the abutments using the dual curing adhesive resin cement Multilink Automix (Ivoclar Vivadent), under a constant pressure of 50 N during a setting period (3 minutes). However, after excess cement was removed, light curing was applied for 20 seconds at each side (buccal - palatal) according to the manufacturer recommendations (Figure 3.4.).

Figure 3.4. Typical group after completion of the bonding procedure.

3.3. Tests and statistics

3.3.1. Dynamic loading test

According to the study outline (Table 3.2.) groups CA, CB, CC, CPB, CPC (n=7) were subjected to thermo-mechanical dynamic loading in a computer-controlled dual-axis chewing simulator (Kausimulator, Willytec, Munich, Germany) for 1,200,000 loading cycles (Figure 3.5), that corresponds to a 5-year clinical fatigue (Kern et al. 1999).

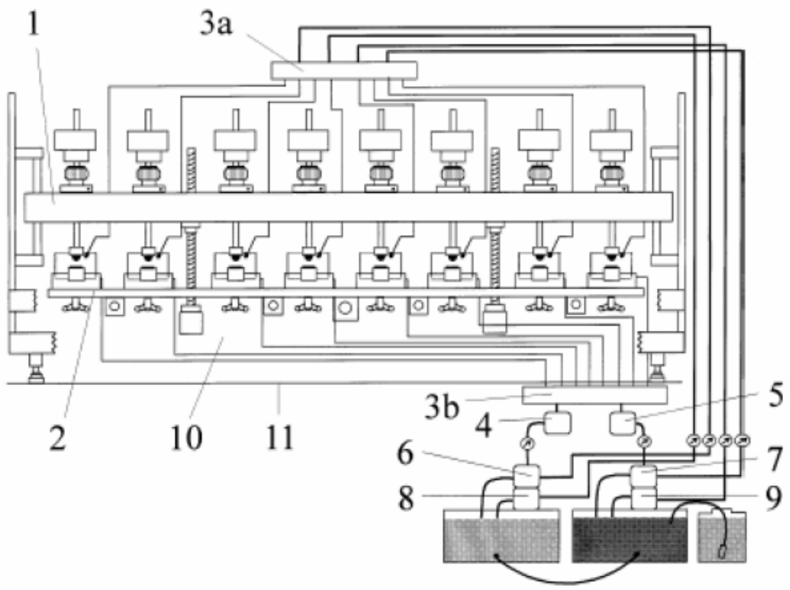

Figure 3.5. Schematic drawing of the dual-axis chewing simulator with eight chambers. *(1) upper crossbeam, (2) lower crossbeam, (3a) water reservoir (in), (3b) water reservoir (out), (4) filter for cold water, (5) filter for warm water, (6) pump for removal of cold water, (7) pump for removal of warm water, (8) pump for application of cold water, (9) pump for application of warm water, (10) motor block, (11) table (Kern et al. 1999).*

During dynamic loading all specimens are allowed to reach a thermal equilibrium between 5°C and 55°C for 60 sec each with an intermediate pause of 12 sec, maintained by a thermostatically controlled liquid circulator (Haake, Karlsruhe, Germany). A loading force of 98 N was applied at an angle of 135° degrees to the horizontal axis, 3 mm below the incisal edge on the palatal 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).

3.3.2. Quasi-static loading test

According to the study outline (Table 3.2.) additional groups SA, SB, SC, SPB, SPC also of seven (7) specimens each and surviving specimens after dynamic loading of groups CA, CB, CC, CPB, CPC, were subjected to quasi-static loading until fracture using a universal testing instrument (Z010/TN2S, Zwick, Ulm, Germany). A semi-spherical loading stamp was centrally positioned in the median plane of the crown between the upper end of the tuberculum and the incisal edge. However, a 1 mm-thick tin foil was placed between loading stamp and crown to achieve homogenous stress distribution. Then, a compressive force was applied at the same angle of 135° degrees to the horizontal axis under stroke control with a cross-head speed of 0.5 mm/min until fracture and fracture strength values (N) achieved were recorded.

3.3.3. Microscopic evaluation

After the quasi-static and the dynamic loading tests, all fractured specimens were ultrasonically cleaned in 96 % alcohol and examined under low power (50 x) stereo-magnification and incident light with the use of an optical microscope (Carl Zeiss, Jena, Germany) and representative photographs were made. 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 possible locations of the fractures. Different fractures types were investigated in order to determine a possible influence the preparation design under loading conditions.

3.3.4. Statistics

As previously described (see chapter 2 and 3.2), the current study examined three different influencing factors regarding the fracture strength of lithium disilicate single implant crowns; a) the preparation depth [0.5/0.7/0.9 mm], b) the preparation mode [milling by the manufacturer/milling using the Celay system] and c) the loading mode [dynamic/quasi-static]. After both dynamic and quasi-static loading tests, fracture strength values were statistically evaluated using the multiple linear regression method.

Regression analysis is used to understand which among the independent variables, such as the preparation mode, the preparation depth and the loading mode, are related to the dependent variable, such as the fracture strength, and to explore the forms of these relationships. Regression analysis may include techniques for modeling and analyzing several variables, when the focus is on the relationship between a dependent variable and one or more independent variables. More specifically, regression analysis estimates the conditional expectation of the dependent variable given the independent variables; that is, the average value of the dependent variable when the independent variables are held fixed (Chan 2004).

4. RESULTS.

4.1. Results of the dynamic loading

With the exception of one specimen in the Group CPB which failed at 300,000 loading cycles due to implant fracture, every specimen in all test groups CA, CB, CC, CPB, CPC survived the 1,200,000 loading cycles in the chewing simulator (Kausimulator, Willytec). Loading cycles (n) achieved for each specimen after dynamic loading are shown in Table 4.1.

Table 4.1. Mean loading cycles (n) achieved after dynamic loading *(No.: number of specimens, S.D.: standard deviation, group codes see Table 3.2.).*

Groups	No.	Mean ± S.D.
CA	**7**	**1,2 10^6±0.00**
CB	**7**	**1,2 10^6±0.00**
CC	**7**	**1,2 10^6±0.00**
CPB	**7**	**1,07 10^6±340.17**
CPC	**7**	**1,2 10^6±0.00**

4.2. Results of the quasi-static loading

Groups SA, SB, SC, SPB, SPC and surviving specimens after dynamic loading of groups CA, CB, CC, CPB, CPC, were subjected to quasi-static loading until fracture using a universal testing instrument (Z010/TN2S, Zwick). Fracture strengths (N) achieved for each specimen after quasi-static loading are shown in Table 4.2.

Table 4.2. Mean fracture strengths (in Newtons) achieved after quasi-static loading *(No.: number of specimens, S.D.: standard deviation, group codes see Table 3.2).*

Groups	No.	Mean (N)	S.D.
SA	7	**383.8**	83.9
CA	7	**403.4**	67.0
SB	7	**294.3**	95.4
CB	7	**374.0**	75.0
SC	7	**331.7**	52.4
CC	7	**372.7**	105.0
SPB	7	**332.4**	79.9
CPB	6	**499.0**	90.7
SPC	7	**380.7**	101.5
CPC	7	**358.1**	53.8
Total	69	371.2	91.5

4.3. Descriptive table and diagram of the study results

A synopsis of the study results is given through Table 4.3 and Figures 4.1. and 4.2.

Table 4.3. Mean, standard deviations, minimum, median, and maximum fracture strengths (in Newtons) of test groups. *(Group codes see Table 3.2.).*

Groups	n	Mean	S.D.	Min	Median	Max	Range
SA	7	**383.8**	83.9	292.0	372.0	544.0	252.0
CA	7	**403.4**	67.0	313.0	389.0	501.0	188.0
SB	7	**294.3**	95.4	198.0	270.0	474.0	276.0
CB	7	**374.0**	75.0	265.0	380.0	481.0	216.0
SC	7	**331.7**	52.4	270.0	332.0	421.0	151.0
CC	7	**372.7**	105.0	251.0	396.0	499.0	248.0
SPB	7	**332.4**	79.9	230.0	299.0	436.0	206.0
CPB	6	**499.0**	90.7	370.0	517.5	613.0	243.0
SPC	7	**380.7**	101.5	255.0	341.0	566.0	311.0
CPC	7	**358.1**	53.8	308.0	327.0	452.0	144.0
Total	69	371.2	91.5	198.0	370.0	613.0	415.0

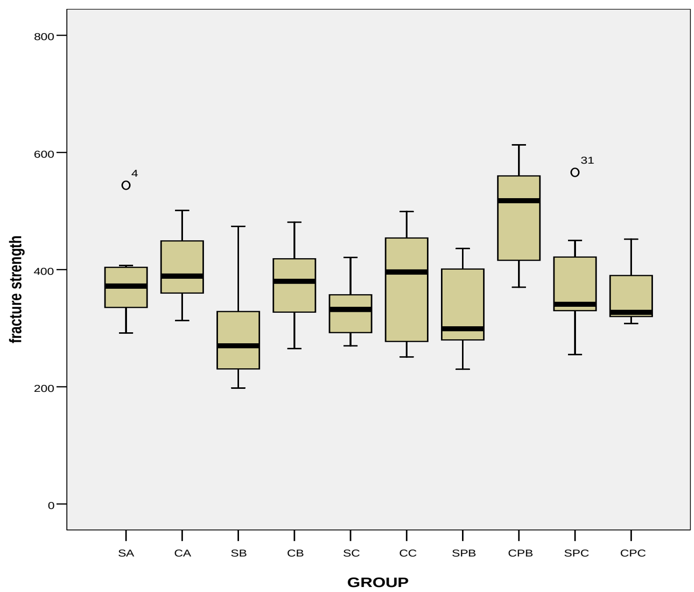

Figure 4.1. Box-plot diagram indicating the load to fracture for all test groups. *(Horizontal lines inside the boxes represent the median values of each group).*

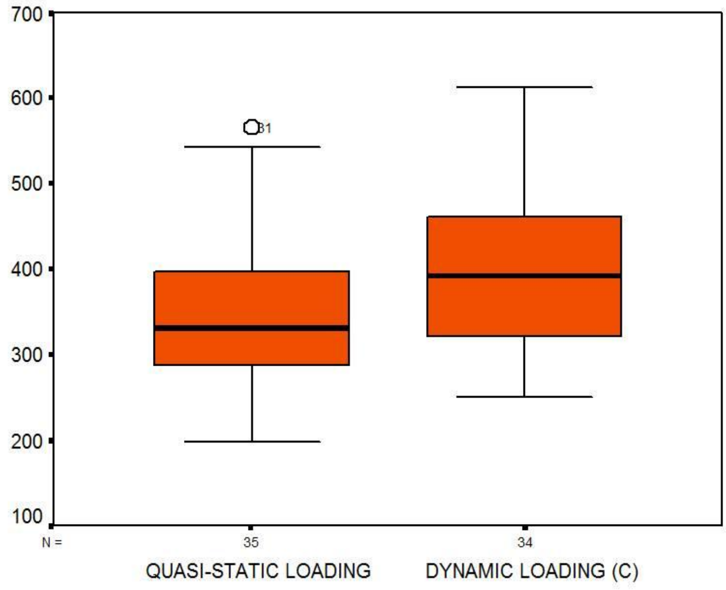

Figure 4.2. Box-plot diagram indicating the overall fracture strength of the test groups subjected to either quasi-static loading or dynamic loading. *(Horizontal lines inside the boxes represent the mean values of each test.)*

4.4. Statistical analysis

The multiple linear regression statistical method followed in the current study uses a linear model that examined the significance of all factors (preparation depth, preparation mode and loading mode) in relation to the fracture strength data (dependent variable). The application of the specific statistical method was validated performing a series of different tests (see appendix). A backward selection method of the independent variables was carried out in order to achieve the final statistical model. For the preparation mode and the loading mode variables, the *"preparation by the manufacturer"* and the *"quasi-static loading"* were entered to the abovementioned model as baselines, respectively. For the variable preparation depth, level labeled as B=0.7 was entered to the linear regression model as the baseline between the two dummy[1] variables labeled as A=0.5 and C=0.9. After the application of the data to the multiple linear regression model the following results were gathered (Table 4.4.). The analytical output of the multiple linear regression is stated at chapter 12 (appendix).

Table 4.4. Multiple linear regression.

Model summary

R	R square	Adjusted R square
0.397	0.158	0.105

ANOVA

Model	Sum of squares	Df	Mean square	F	P value
Regression	89953.766	4	22488.441	3.001	0.025*
Residual	479587.481	64	7493.554		
Total	569541.246	68			

*Predictors: (Constant), loading mode, preparation mode, preparation depth (A=0.5, C=0.9)

Variables	Category	Regression Coefficient b	SE(b)	P value
Loading mode	Quasi-Static	Baseline		
	Dynamic	54.60	20.85	0.01
Preparation mode	Manufacturer	Baseline		
	Celay system	46.67	23.36	0.05
Preparation depth	B	Baseline		
	A	44.56	30.65	0.15
	C	-11.387	23.36	0.628

Statistical analysis revealed that the mean fracture strength of the lithium disilicate implant crowns over manually prepared zirconia abutments was slightly increased than the ones over zirconia abutments prepared by the manufacturer (p=0.05). Thus,

[1] Categorical variable that represents subgroups of the study specimens in regression analysis.

"*preparation mode*" as influencing factor seems not to play a statistically important role concerning the fracture strength outcome. In addition, despite that level A (=0.5 mm) resulted to an increase and, conversely, level C (=0.9 mm) to a decrease of the fracture strength, both data showed no statistical significance (p=0.15 and p=0.628, respectively). Therefore, it can be assumed that using zirconia implant abutments, the *"preparation depth"* had no influence on the fracture strength of the lithium disilicate implant crowns after dynamic and/or quasi-static loading. Last but not least, the mean fracture strength of the lithium disilicate implant crowns was found statistical significantly increased after dynamic loading (p=0.01<5%) than the ones subjected to quasi-static loading only. From the observation of the beta values, it can also be concluded that the variable "loading mode" had a major effect on the fracture strength of the study specimens (beta= 0.300), followed by the "preparation mode" (beta= 0.251). Finally, it is notable that the variability of the fracture strength values was almost 16% (R square=0.158), leading to the hypothesis that there might be more influencing parameters than the ones examined in the current study.

4.5. Fracture mode

Fracture patterns were recorded and evaluated after failure in every specimen. All zirconia abutments had a typical failure mode. More specifically fractures occurred at the most-tapered part of the abutment internal to the implant hex as shown in Figure 4.3. No screw bending or loosening were observed during both static and dynamic loading. However; during dynamic loading there was observed one implant fracture.

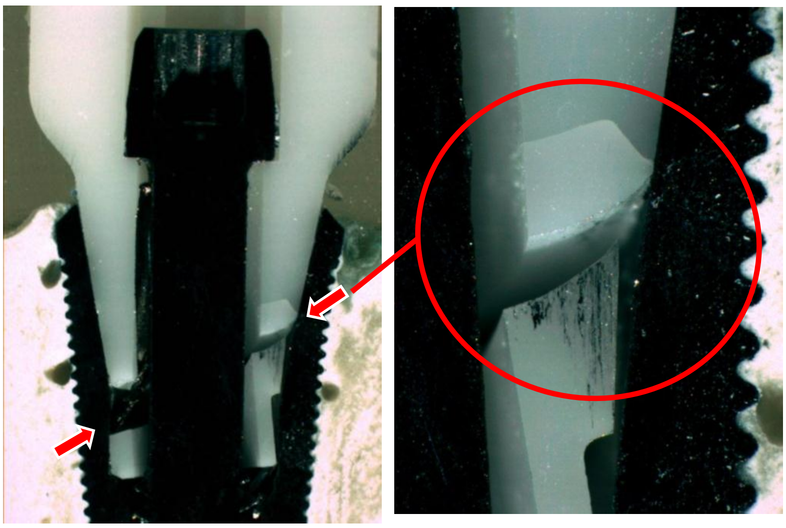

Figure 4.3. Microscopic image (x12.5) of a vertically sectioned specimen at the implant/abutment joint.

5. DISCUSSION

5.1. Discussion of the methods

5.1.1. Abutment preparation

In several in-vivo and in-vitro studies, circumferential shoulder preparations of 1.0 to 1.5 mm were routinely used for the fabrication of all-ceramic crowns (Attia and Kern 2004a, Pröbster 1992, Beschnidt 1998). The study design of the current study followed the manufacturer’s (Ivoclar) preparation guidelines in conjunction with lithium disilicate crowns. Therefore the zirconia abutments in all groups were prepared with a circumferential chamfer preparation, however; sharp transitions, inner angles and feather edges were avoided. The preparation finish line can influence crown marginal adaptation (Lin et al. 1998), but using the chamfer preparation in comparison to the shoulder one either seems to improve the marginal fit (Pera et al. 1994) or not to present a significant difference (Shearer et al. 1996). Moreover, the chamfer preparation finish line may help adhesive cement, used in this study, to escape during seating and therefore to improve marginal adaptation (Gavelis et al. 1981).

Furthermore three different preparation depths of 0.5 or 0.7 or 0.9 mm were also performed and tested according to the protocol (see Table 3.2. in chapter 3.2.1; study outline). The specific preparation depths, which are even narrower than the ones used for the preparation of natural teeth, were considered enough for implant abutment preparations. In addition, all abutments received a 6-degree tapered angle axial preparation which was also commonly used in laboratory studies (Mc Cormick et al. 1993, Strub and Beschnidt 1998, Attia and Kern 2004a).

Regarding the preparation mode, all zirconia abutments were prepared by appropriate milling following given specifications either by the manufacturer (Astra Tech) or by an experienced dental technician. Using prefabricated zirconia blocks, manual abutment milling with a corresponding system such as the Celay system (Mikrona) could be very beneficial in order to achieve identical abutment dimensions relative to a preset master die. However, zirconia grinding or milling might induce surface flaws or microcracks which might influence the mechanical properties of the material (Luthardt et al. 2004). It has been confirmed that the aforementioned surface treatment generally triggers T-M transformation which negatively influences the mechanical properties of the material after coarse grinding (Wang et al. 2008, Rekow and Thompson 2005). A stress-free abutment preparation under water cooling using a fine-grained cutting diamonds, as followed in the current study, may decrease the critical flow size and increase the surface compressive layer which provides which improves strength (Kosmac et al. 1999, Kosmac et al. 2000).

5.1.2. Crown fabrication

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 the esthetic performance in order to provide space for the veneering materials.

Metal implant abutments may provide even increased dimensions of preparation because there is no endodontic limitation such as during the preparation of the natural teeth, however; the all-ceramic crown restoration might fail to adequately mask the underlying metal-shade abutment. In the present study, lithium disilicate implant crowns achieved high fracture strengths when bonded over zirconia abutments which were prepared to the same extend as described above for the natural teeth. In addition, due to the pleasing abutment shade of the zirconia the combination of such crowns could be gathered advantageous from the esthetic point of view.

The thickness of crowns was standardized during designing both the shape and the dimension of the crown~~s~~ by duplicating the initial wax-up with a silicon index. Moreover, the thickness of lithium disilicate crowns (e.max Press, Ivoclar Vivadent) during the laboratory fabrication, was verified with the use of a caliper followed, if needed, by the removal of superficial ceramic mass using a porcelain finishing stone (Attia and Kern 2004a, Attia and Kern 2004b).

5.1.3. Crown cementation

Adhesive cementation is recommended for all-ceramic restorations to enhance their fracture strength. Therefore, a dual curing resin cement (Multilink, Ivoclar Vivadent) was used for the purposes of this study. Resin cements may be used for the cementation of zirconia based restorations, but is not mandatory (Raigrodski 2004), as the bonding of resin with zirconia ceramic is difficult to be achieved (Derand and Derand 2000). 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 (Kern and Wegner 1998). Only bonding systems that contain a special adhesive monomer have been found to provide an acceptable, high strength, stable bond to air-abraded Y-TZP. However; the retention can be increased for instance by air-abrasion with 50 µm alumina particles (Wegner and Kern 2000). In this study a phosphoric / phosphonic acid reagent (Metal/Zirconia Primer, Ivoclar) was used to form low-soluble, stable phosphates / phosphonates with the zirconia abutment (Kern et. al. 2009, Lehmann and Kern 2009). The active reagent of the primer is a methacrylate monomer which has one phosphonic acid group. Similarly to silane on silicate ceramic, chemical bonding is made possible and the zirconium oxide surface can be wetted with the luting composite. According to the manufacturer, this conditioning is stable enough to withstand the stress of thermocycling.

On the other hand, bonding to silica based ceramic may be very effective by using a resin luting agent after hydrofluoric 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 (Blatz et al. 2003, Klosa et al. 2009).

5.1.4. Tests

In order to accept a dental material or restoration design for clinical use, 5-year clinical results should be available. Moreover, a prosthetic restorative system can be successfully considered if it demonstrates a survival rate of 95% after 5 years and 85% after 10 years (Strub 1992, Pröbster 1996). However, clinical studies that could accurately evaluate the biomechanical behavior or the clinical success of materials and restorations need increased costs and time (Kern et al. 1999). Therefore, in vitro tests which have the potential to simulate clinical conditions, as in the present study, could provide reliable outcomes and save expenses and evaluation time.

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 as clinical relevant testing conditions than monotonic loading (Kelly 1999). The wear of a restoration after 2.4 to 2.5 x 10^5 masticatory cycles in the chewing simulator corresponds to 1 year of clinical service (De Long 1985). In the present study, the chewing simulator was set to perform 1.2 x 10^6 masticatory cycles, simulating 5 years of clinical service (Kern et al. 1999).

In addition, the specific chewing simulator used in this study was developed to reproduce testing conditions under controlled moisture and thermocycling. 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 (Chevalier 2006).

The direction of the loading forces may significantly influence the fracture strength of all ceramic restorations (Koutayas et al. 2000). In this study, loading forces were applied under 135°, regarding the longitudinal axis, and with the use of a 6-mm ceramic ball, positioned in the midline of the crown, to imitate teeth contact correlation in the anterior region during physiological incisive movements. Antagonist steatite ceramic presenting a similar to enamel hardness (Vicker's scale) and was gathered as a suitable substitute material for enamel in wear tests (Kelly 1999). The magnitude, duration and frequency of the loading force applied in the chewing simulator were comparable to the values reported in the literature (Krejci et al. 1990). The applied effective loading force of 100N was between the limits of the maximum physiological biting forces in the anterior region, however; such loading force magnitude has never tested before for so prolonged time (Koutayas et al. 2000, Butz et al. 2005, Att et al. 2006a, b, Steiner et al. 2009).

Furthermore, dynamic loading using chewing simulators have been proved useful in mimicking human oral conditions and application of physiological chewing forces and therefore in testing such restorations under fatigue conditions (i.e. mechanical cyclic loading, thermocycling in aqueous environment) (Steiner et al. 2009).

In-vitro evaluation under quasi-static loading of dentin-bonded all-ceramic crowns under compressive load might give some indication about the clinical durability of these restorations. Several factors influence the fracture loads of all-ceramic crowns, such as the microstructure of the ceramic material (Della Bona et al. 2002, Oh et al. 2000), the fabrication technique, the final surface finishing of the crowns (Chen et al. 1999) and the luting method (Burke and Watts, 1998, Malament and Socransky 2001). Other important factors are the 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 (Kelly 1999, Yoshinari and Derand 1994). In the present these factors were standardized as close as possible to the clinical conditions.

Finally, in order to provide meaningful results regarding the durability of such implant crowns quasi-static loading was carried out after mechanical testing in the chewing simulator under dry and wet conditions within a thermal range (Attia and Kern 2004b, Ohyama et al. 1999, Yoshinari and Derand 1994).

5.2. Discussion of the results

5.2.1. Dynamic loading

In the present study, half of the specimens of each test group were exposed to the chewing simulator before the fracture strength test was performed. All specimens, with one exception in group CPB (see Table 4.1.) survived the dynamic loading test (1.2×10^6 loading cycles). The specific specimen was disassembled and evaluated with the use of an optical stereoscope and scanning electron microscopy (SEM). The observed failure at the implant level (implant fracture) under 100 N was considered as manufacturing liability since it was demonstrated that in order for an implant to be fractured a static force of at least 900N should be applied (Mitsias et al. 2010). All other components of the specimen, including the zirconia abutment and the all-ceramic crown, were found in perfect condition without any fractures. Therefore, the specific specimen was not taken into the statistical evaluation for group CPB.

According to the literature, an average of approximately 250,000 cycles in a chewing simulator corresponds to one year of clinical service (DeLong and Douglas 1991, Kreijci and Lutz 1990). Therefore mean 1.2×10^6 dynamic loading cycles achieved by the implant crowns in all study groups (Table 4.1) without fracturing, corresponded to a 5-year service time. Comparing the physiological bite forces that may range from 10 to 120 N during chewing of food or swallowing (De Boever et al. 1978, Kohyama et al. 2004) to the loading forces applied during dynamic loading, it was demonstrated that for the desired loading force of 100 N, the expected relative mean overload found up to 8.1% which is within the aforementioned physiological range (Steiner et al. 2009). Therefore, it could be presumed that for an effective loading force of 100 N, adhesively cemented lithium disilicate single implant crowns over zirconia abutments could withstand maximum physiological biting forces on a long-term basis.

Additionally, regardless the preparation mode or depth of the zirconia abutments, it was found that dynamic loading in the artificial mouth increased the fracture strength of all groups with the exception of group CPC. This finding, illustrated in Figure 4.2., cannot be supported by evidence based scientific data, however; it has been also previously observed after a similar dynamic loading test concerning all-ceramic crowns placed on endodontically treated teeth (Friedel and Kern 2006) and might indicate an advantageous behavior of the adhesively cemented crowns under loading which should be evaluated through further in-vitro studies.

Several in-vitro studies explored the feasibility of zirconia abutments to withstand functional loading in the oral environment (Butz et al. 2005, Att et al. 2006a, Att et al. 2006b). These studies utilized similar methods where 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 all these studies noted high survival rates of 100% after an equivalent of 5-year chewing simulation without any screw loosening in agreement to the present study in which even a twofold loading force was used (100 N). Therefore, 5-year in-vitro data support the use of zirconia abutments in the anterior regions. The latter could be also verified by a recent systematic review that identified a 0.2% clinical failure rate of ceramic abutments [per 100 abutment years, (95% CI: 0.02–1.3%)] and a 5-year survival rate of 99.1% (95% CI: 93.8–99.9%) (Sailer et al. 2009). Additionally, it was concluded that, ceramic abutments when supporting implant crowns can be considered as a valid alternative to the metal ones as they exhibit similar survival (97.4%) and complication rates up to 3 years.

5.2.2. Quasi-static loading

Mean fracture strengths after every different loading test (dynamic and quasi-static loading or only quasi-static loading) are shown in Table 4.2. Linear regression model (R square =0.158) did not reveal any statistical differences between the groups meaning that the study variables (preparation mode and preparation depth) did not influence the strength characteristics of the specific implant crowns. Nevertheless statistics pointed out that the variable “loading mode” had a major effect on the fracture strength of the study specimens (beta= 0.300). The fracture strength improvement after dynamic loading might be attributed to strengthening transformation phenomena of zirconia abutment under loading or to the achieved stress relaxation of the complex implant/abutment/crown due to the cementation mean or/and the physical properties of the metal screw or even the enhancement of the adaptation due to a possible micro-abrasion between the abutment and the implant platform. Fracture strengths in all study groups varied between 294 and 499 N and therefore could be considered to be within or above the limits of the maximum physiological forces generated in the maxilla anterior region (Killiaridis et al. 1993).

Previous in-vitro studies that examined the mechanical stability of implant crowns placed on zirconia abutments by assessing the mean fracture strengths (in N) are included in Table 5.1. The preparation depth of the zirconia abutments in all studies stated above range between 0.5 to 1.0 mm while the preparation mode was performed either manual or it was used the original abutment. Regarding the specific zirconia abutment (ZirDesign, AstraTech), Mitsias et al. (2010) found a higher fracture strength value of 690 N which might be explained by the use of a stiffer crown material such as a non-precious alloy than in the present study. Adatia et al. in a recent study (2009) also tested the same abutment type using two different preparations depths of 0.5 and 1.0 mm, however; clinical relevance of the study might be doubtful because of the use of implant analogues instead of original implants and the lack of a crown restoration. Nevertheless, all laboratory studies (Table 5.1.) demonstrated mean fracture strengths beyond the clinical acceptance; therefore zirconia abutments seem to be a promising treatment option (Koutayas et al. 2009).

Table 5.1. Comparison of current in-vitro studies that examined the fracture strength of single anterior implant crowns placed on zirconia abutments *[N.R.: not referred, (*): dynamic loading followed by static loading of the survived specimens]*.

Author, Year	**Implant**			**Zirconia Abutment**	**Preparation, Depth (mm)**	**Preparation Mode**	**Ceramic Crown**	**Loading Direction**	**Loading Test(s)**	**Mean F.S. (N) ±SD**
	Name	**D(mm)**	**L(mm)**							
Yildirim et al., 2004	Brånemark External Analog, NobelBiocare	N.R.	N.R.	Wohlwend Innovative	Chamfer, 1.0	Celay system	Leucite reinforced ceramic	30º	Static	737±245
Mitsias et al., 2009	OsseoSpeed, AstraTech	4.5	15	ZirDesign	Chamfer, 0.5	Manual Milling	Non-precious alloy	30º	Static	690±430
Butz et al., 2005	Osseotite (external), Biomet 3i	4.0	13	ZiReal	Chamfer, 0.5	Manufacturer	Non-precious alloy	50º	Dynamic*	281±N.R.
Att et al., 2006(a)	Replace Select, NobelBiocare	4.3	15	Esthetic Zirconia Abutment	Chamfer, 0.5	Manual milling	Densely sintered Alumina	50º	Dynamic*	470±152
Att et al., 2006(b)	Replace Select, NobelBiocare	4.3	15	Esthetic Zirconia Abutment	Chamfer, 0.5	Manual milling	Densely sintered Zirconia	50º	Dynamic*	593±292
Aramouni et al., 2008	Certain, Biomet 3i	4.0	13	ZiReal	Chamfer, 1.0	Manual milling	Lithium disilicate	45º	Static	793±123
Adatia et al., 2009	OsseoSpeed Analogue, AstraTec	N.R.	N.R.	ZirDesign	Chamfer, 0.5	Manual Milling	Without crown	30º	Static	576±140
					Chamfer, 1.0	Manual Milling				547±139
CURRENT STUDY	OsseoSpeed, AstraTech	4.5	15	ZirDesign	Chamfer, 0.5	Manufacturer	Lithium disilicate	30º	Dynamic*	403±67
					Chamfer, 0.7	Manufacturer				374±75
					Chamfer, 0.7	Celay system				372±105
					Chamfer, 0.9	Manufacturer				499±91
					Chamfer, 0.9	Celay system				358±54

For the given angle of load application of 30º, mean fracture strengths in the present study were found within the range of the fracture loads (≥500 N) described in a systematic review, with respect to either the abutment and the restoration materials or the internal implant–abutment connection (Figure 5.1.). Finally, in order to achieve better direct comparisons and export high clinical relevant data, standardization of future laboratory tests that evaluate the strength of abutments is needed (Hobkirk and Wiskott 2009).

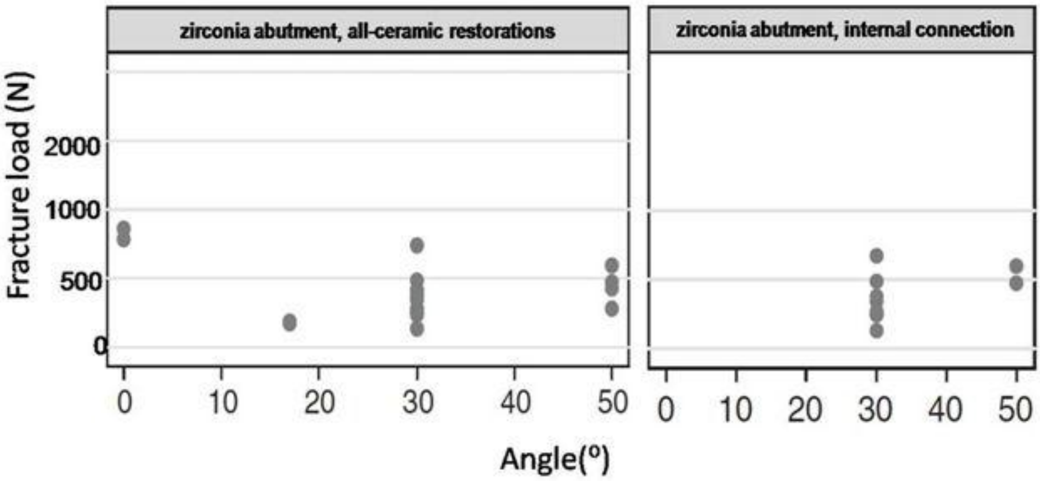

Fig. 5.1. Fracture load (N) with respect to abutment/restoration material [left] and zirconia abutment material, internal implant–abutment connection [right] and angle of load application (°) *[Source:Sailer et al. 2009]*.

5.2.3. Failure mode

Study findings in the majority of the specimens revealed that fractures were located at the cervical aspect of the abutments at 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. The internal cone of the particular zirconia abutment seems to be a high loaded component that receives torque and stress concentrations. Crack propagation seems to be related to the magnitude and the application point of the loading forces and the fulcrum location *(=pivot where the lever moves)*. Therefore induced loading forces (≥ 294 N) when applied under an angle of 30º (or 150º) may cause levering effects such as in a second class lever. 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 (Fig. 5.2.). For this reason, the output load is applied in an area where the internal cone of the abutment originally has thinner walls which obviously cannot withstand the specific loading. Moreover, the fracture strength depends on the extension of the crown margin relative to the location of the screw head (Tripodakis et al. 1995). Implant design with internal connection, such as the ones used in the current study, may increase this abovementioned extension, however; it was illustrated that internal connection of abutments tends to be beneficial both in laboratory and in clinical studies (Sailer et al. 2009).

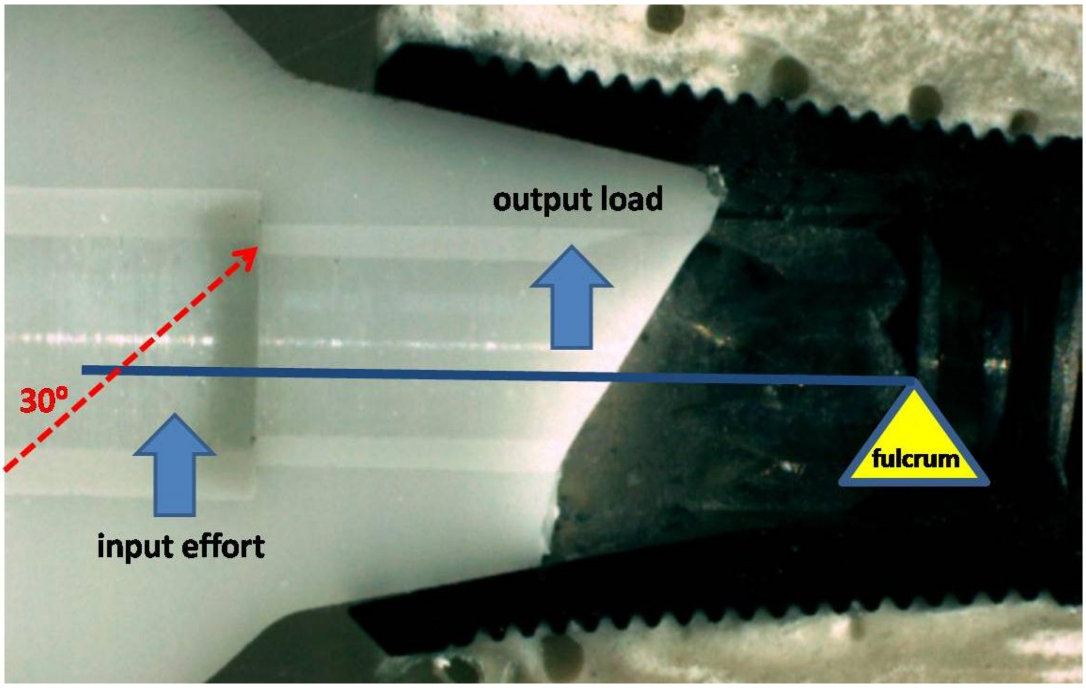

Fig. 5.2. Second class levering effects within the internal connection of the zirconia abutment (Red dashed line represents the loading direction).

6. CONCLUSIONS

Within the limitations of the present study the following conclusions can be drawn:

1. The preparation mode (CAD/CAM machining or manually controlled) of customized zirconia implant abutments seems not to influence the fracture strength of adhesively cemented single implant lithium disilicate crowns and their abutments.
2. A zirconia ceramic abutment preparation depth up to 0.9 mm circumferentially had no negative effect on the fracture strength of adhesively cemented single implant lithium disilicate crowns and their abutments.
3. Dynamic loading may improve fracture strength and therefore the durability of adhesively cemented single implant lithium disilicate crowns placed on zirconia abutments.
4. Failure of single implant lithium disilicate crowns placed on zirconia abutments was located at the level of the implant/abutment internal connection. However; fractures occurred under higher forces than the expected maximum physiological chewing forces.

7. SUMMARY

Zirconia implant abutments offer enhanced esthetics and promote biological sealing; however, the effect of mechanical processing due to preparation has not been investigated under functional loading. The purpose of the study was to evaluate the influence of the zirconia abutment preparation depth and preparation mode on the survival rate, the fracture strength and fracture mode of all-ceramic single implant crowns.

Seventy single implant-supported lithium disilicate glass-ceramic crowns (IPS e.max Press, Ivoclar Vivadent) were adhesively cemented (Multilink Automix, Ivoclar Vivadent) onto zirconia abutments (ZirDesign, Astra Tech) using implants with a diameter of 4.5 mm and a length of 15.0 mm (Osseospeed, Astra Tech). They replaced a maxillary central incisor (11.0 mm in height and 8.0 mm in width). Lithium disilicate implant crowns were divided into 5 study groups (n=14) according to the abutment preparation depth [(A: control) 0.5, (B:) 0.7, (C:) 0.9 mm, and preparation mode [milling by the manufacturer, (P:) milling by the Celay System (Mikrona)]. Subgroups (n=7) were subjected to dynamic loading (C) at 135° with 98N in a thermomechanical chewing simulator (Kausimulator, Willytech) up to 1.2×10^6 loading cycles; followed by quasi-static loading until fracture.

All specimens survived dynamic loading except one (in group CPB) that fractured early and was considered as manufacturer's mal-production. Additional subgroups (n=7) were subjected to quasi-static loading (S) at 135° in a universal testing machine (0.5 mm/min, Z010/TN2S, Zwick). Mean fracture strengths (N) were: Group SA: 384±84; Group CA: 403±67; Group SB: 294±95; Group CB: 374±75; Group SC: 332±52; Group CC: 373±105; Group SPB: 332±80; Group CPB: 499±91; Group SPC: 380±101; Group CPC: 358±54. Statistical analysis using multiple linear regression showed that both the preparation depth and mode had no influence on the fracture strength of the implant crowns ($p>0.05$), however; fracture strength increased statistically significantly after dynamic loading ($p=0.01$).

Adhesively luted single implant lithium disilicate crowns placed on zirconia abutments have the potential to withstand physiological maximal incisive biting forces for more than 5 years of simulated fatigue. Manually controlled circumferential chamfer zirconia abutment preparation had no effect after 5 years simulated dynamic loading. However, single implant lithium disilicate crowns placed on zirconia abutments seem to increased fracture strength after dynamic loading.

8. ZUSAMMENFASSUNG

Zirkonimplantatabutments erfüllen erhöhte ästhetische Ansprüche und verbesserte biologische Integration. Die Auswirkung mechanischer Belastungen in Abhängigkeit unterschiedlicher präparativer Bearbeitung wurde noch nicht unter funktioneller Belastung untersucht. In dieser Studie wurde die Überlebensrate, Bruchfestigkeit und Art des Versagens von Lithiumdisilikat-Glaskeramikkronen (IPS e.max Press, Ivoclar Vivadent) auf Zirkonoxidkeramikabutments (Astra Tech AB) nach künstlicher Alterung im Kausimulator überprüft.

Siebzig einzelne implantatgetragene Lithiumdisilikat-Glaskeramikkronen (IPS e.maxPress, Ivoclar Vivadent) wurden adhäsiv (Multilink Automix, Ivoclar Vivadent) auf Zirkonoxidkeramikabutments (ZirDesign, Astra Tech) befestigt. Die Implantate (Osseospeed, Astra Tech) hatten einen Durchmesser von 4,5 mm und eine Länge von 15 mm. Die Implantatkronen ersetzen zentrale Oberkiefer-Frontzähne mit einer Höhe von 11 mm und einer Breite von 8 mm. Die Implantatkronen aus Lithiumdisilikat wurden unter Berücksichtigung der Abutmentpräparationstiefe [A (Kontrollgruppe): 0.5, B: 0.7, C: 0.9 mm und Präparationsart - Fräsung durch den Hersteller, P: Fräsung mit Hilfe des Celay-Systems (Mikrona)] in 5 Gruppen (n=14) eingeteilt. Untergruppen (n=7) wurden für $1{,}2 \times 10^6$ Zyklen in einem Kausimulator (Willytech) dynamischen Belastungen (C) unter einem Winkel von 135° mit 98 N ausgesetzt. Anschließend folgte eine quasi-statische Belastung bis zum Bruch.

Alle Proben überlebten die dynamischen Belastungen außer eine aus der Gruppe CPB, die vorzeitig gebrochen war; dieses Versagen wurde einem Herstellungsfehler zugeordnet. Zusätzliche Untergruppen (n=7) wurden quasi-statischen Belastungen (S) in einem Winkel von 135° in einer universellen Testmaschine (0,5 mm/min, Z010/TN2S, Zwick) ausgesetzt.

Die resultierenden Bruchfestigkeiten (N) waren: Gruppe SA: 384.8±83.9; Gruppe CA: 403.4±67.0; Gruppe SB: 294.3±95.4; Gruppe CB: 374.0±75.0; Gruppe SC: 331.7±52.4; Gruppe CC: 372.7±105.0; Gruppe SPB: 332.4±79.9; Gruppe CPB: 499.0±90.7; Gruppe SPC: 380.7±101.5; Gruppe CPC: 358.1±53.6. Die statistische Analyse mittels multipler linearer Regression zeigte, dass weder die Präparationsart noch die Präparationstiefe einen signifikanten Einfluss auf die Bruchfestigkeit der Implantatkronen ($p>0{,}05$) hatte: hingegen war die Bruchfestigkeit nach dynamischer Belastung statistisch signifikant erhöht ($p=0{,}01$).

Adhäsiv befestigte Lithiumdisilikat-Einzelkronen auf Zirkonabutments haben das Potential, maximalen, physiologischen Kaukräfte für mehr als 5 Jahre simulierter Abnutzung zu widerstehen. Manuell durchgeführte umlaufende Stufenpräparationen hatten keine Auswirkung nach 5 Jahren auf die Belastbarkeitund nach dynamischer Belastung. Lithiumdisilikatkronen auf Einzelimplantaten mit Zirkonabutments wiesen nach dynamischer Belastung eine Erhöhung ihre Bruchfestigkeit auf.

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The present thesis is dedicated to my parents Emmanuel and Evangelia for their endless support throughout the journey of my life and to my precious grandfather who recently passed away and had me more than a son.

12. APPENDIX

Detailed data of the results

Table 12.1. Loading cycles (n) of all specimens after dynamic loading.

Groups	Specimen No.						
	1	2	3	4	5	6	7
CA	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$
CB	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$
CC	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$
CPB	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^5$
CPC	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$	$1.2x10^6$

Table 12.2. Fracture strengths (N) of all specimens after quasi-static loading.

Groups	Specimen No.						
	1	2	3	4	5	6	7
SA	300	407	401	544	372	371	292
CA	439	313	377	343	462	389	501
SB	335	474	198	200	322	270	261
CB	398	439	353	380	265	302	481
SC	302	270	366	332	421	348	283
CC	413	269	251	499	495	396	286
SPB	298	416	299	262	386	230	436
CPB	613	527	416	508	560	370	-
SPC	393	341	566	450	330	330	255
CPC	318	452	327	322	383	308	397

Testing the assumptions of multiple linear regression

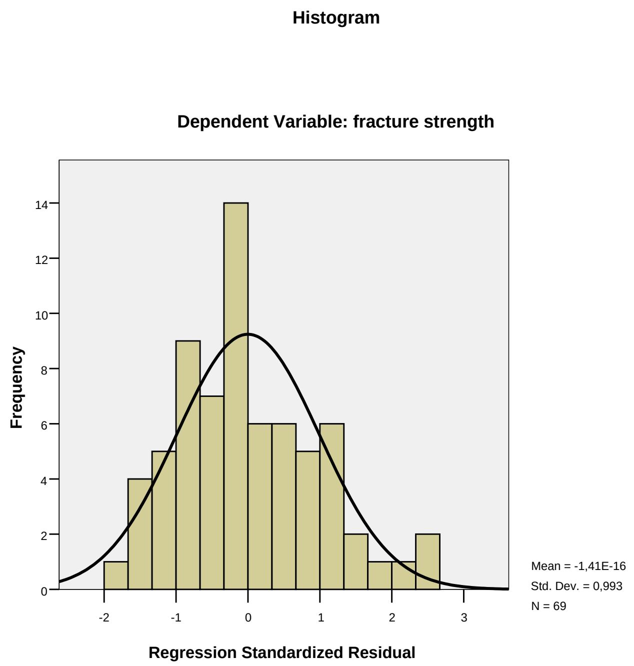

Figure 12.1. Test of independence data (Durbin-Watson Index= 1.705).

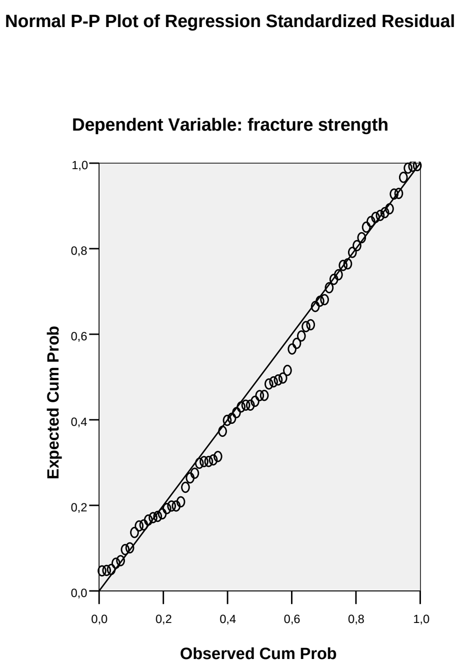

Figure 12.2. Test of normality of residuals.

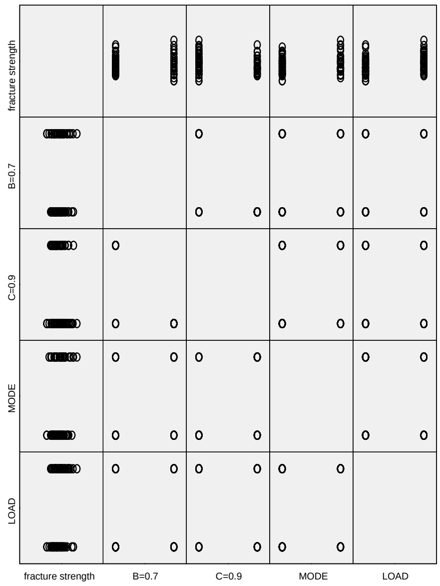

Figure 12.3. Test of linearity.

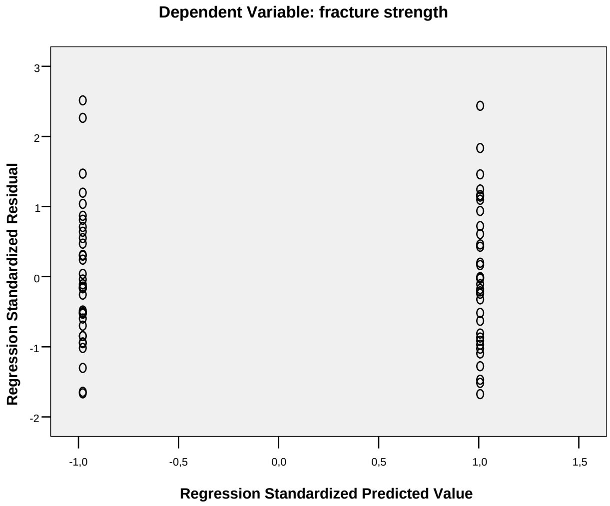

Figure 12.4. Test of homoscedasticity.

Regression

Table 12.3. Descriptive statistics.

	Mean	Std. Deviation	N
fracture strength	371,16	91,518	69
A=0.5	,20	,405	69
depth C=0.9	,41	,495	69
MODE	,39	,492	69
LOAD	,49	,504	69

Table 12.4. Correlations.

		fracture strength	A=0.5	depth C=0.9	MODE	LOAD
Pearson Correlation	fracture strength	1,000	,124	-,094	,154	,297
	A=0.5	,124	1,000	-,417	-,405	,007
	depth C=0.9	-,094	-,417	1,000	,184	,012
	MODE	,154	-,405	,184	1,000	-,018
	LOAD	,297	,007	,012	-,018	1,000
Sig. (1-tailed)	fracture strength	.	,156	,221	,103	,007
	A=0.5	,156	.	,000	,000	,476
	depth C=0.9	,221	,000	.	,065	,461
	MODE	,103	,000	,065	.	,441
	LOAD	,007	,476	,461	,441	.
N	fracture strength	69	69	69	69	69
	A=0.5	69	69	69	69	69
	depth C=0.9	69	69	69	69	69
	MODE	69	69	69	69	69
	LOAD	69	69	69	69	69

Table 12.5. Model summary.

					Change Statistics					
Model	R	R Square	Adjusted R Square	Std. Error of the Estimate	R Square Change	F Change	df1	df2	Sig. F Change	Durbin-Watson
1	,397[a]	,158	,105	86,565	,158	3,001	4	64	,025	
2	,393[b]	,155	,116	86,056	-,003	,238	1	64	,628	1,778

a. Predictors: (Constant), LOAD, A=0.5, MODE, depth C=0.9

b. Predictors: (Constant), LOAD, A=0.5, MODE

c. Dependent Variable: fracture strength

Table 12.6. Variables entered/removed

Model	Variables Entered	Variables Removed	Method
1	LOAD, A=0.5, MODE, depth C=0.9[a]	.	Enter
2	.	depth C=0.9	Backward (criterion: Probability of F-to-remove >= ,100).

a. All requested variables entered.

b. Dependent Variable: fracture strength

Table 12.7. ANOVA.

Model		Sum of Squares	df	Mean Square	F	Sig.
1	Regression	89953,766	4	22488,441	3,001	,025[a]
	Residual	479587,5	64	7493,554		
	Total	569541,2	68			
2	Regression	88172,518	3	29390,839	3,969	,012[b]
	Residual	481368,7	65	7405,673		
	Total	569541,2	68			

a. Predictors: (Constant), LOAD, A=0.5, MODE, depth C=0.9

b. Predictors: (Constant), LOAD, A=0.5, MODE

c. Dependent Variable: fracture strength

Table 12.8. Coefficients.

Model		Unstandardized Coefficients		Standardized Coefficients			95% Confidence Interval for B		Correlations			Collinearity Statistics	
		B	Std. Error	Beta	t	Sig.	Lower Bound	Upper Bound	Zero-order	Partial	Part	Tolerance	VIF
1	(Constant)	321,574	22,552		14,259	,000	276,522	366,627					
	A=0.5	44,556	30,647	,197	1,454	,151	-16,669	105,781	,124	,179	,167	,715	1,399
	depth C=0.9	-11,387	23,356	-,062	-,488	,628	-58,046	35,272	-,094	-,061	-,056	,826	1,211
	MODE	46,673	23,356	,251	1,998	,050	,014	93,332	,154	,242	,229	,836	1,196
	LOAD	54,596	20,851	,300	2,618	,011	12,941	96,250	,297	,311	,300	,999	1,001
2	(Constant)	315,966	19,284		16,385	,000	277,454	354,479					
	A=0.5	50,250	28,169	,222	1,784	,079	-6,006	106,506	,124	,216	,203	,836	1,196
	MODE	46,459	23,215	,250	2,001	,050	,096	92,822	,154	,241	,228	,836	1,196
	LOAD	54,424	20,725	,299	2,626	,011	13,033	95,816	,297	,310	,299	1,000	1,000

a. Dependent Variable: fracture strength

Table 12.9. Coefficient correlations.

Model			LOAD	A=0.5	MODE	depth C=0.9
1	Correlations	LOAD	1,000	-,006	,017	-,017
		A=0.5	-,006	1,000	,367	,381
		MODE	,017	,367	1,000	-,019
		depth C=0.9	-,017	,381	-,019	1,000
	Covariances	LOAD	434,766	-4,102	8,203	-8,203
		A=0.5	-4,102	939,258	262,500	272,754
		MODE	8,203	262,500	545,508	-10,254
		depth C=0.9	-8,203	272,754	-10,254	545,508
2	Correlations	LOAD	1,000	,000	,017	
		A=0.5	,000	1,000	,404	
		MODE	,017	,404	1,000	
	Covariances	LOAD	429,545	6,16E-014	7,955	
		A=0.5	6,20E-014	793,465	264,488	
		MODE	7,955	264,488	538,920	

a. Dependent Variable: fracture strength

Table 12.10. Excluded variables.

Model		Beta In	t	Sig.	Partial Correlation	Collinearity Statistics		
						Tolerance	VIF	Minimum Tolerance
2	depth C=0.9	-,062[a]	-,488	,628	-,061	,826	1,211	,715

a. Predictors in the Model: (Constant), LOAD, A=0.5, MODE

b. Dependent Variable: fracture strength

Table 12.11. Residuals statistics.

	Minimum	Maximum	Mean	Std. Deviation	N
Predicted Value	315,97	420,64	371,16	36,009	69
Residual	-132,425	203,575	,000	84,137	69
Std. Predicted Value	-1,533	1,374	,000	1,000	69
Std. Residual	-1,539	2,366	,000	,978	69

a. Dependent Variable: fracture strength