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Alumina reinforced zirconia implants: Effects of cyclic loading and abutment modification on fracture resistance

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ABSTRACT

Objective. The aim of the study was to evaluate the thermomechanical behavior of alumina-toughened zirconia (ATZ) oral implants in the artificial mouth and the fracture resistance (fracture load and bending moment) in a subsequent static fracture load test. The effects of abutment modification and different cyclic loadings were evaluated.

Methods. A total of 48 implants were used. 24 implants were left as machined (Group A), and 24 implants were shape modified at the abutment (Group B). Groups were divided into three subgroups composed of 8 samples each (A1/B1: no cyclic loading; A2/B2: 1.2 million cycles; A3/B3: 5 million cycles). Subsequently, all implants were statically loaded to the point of fracture.

Results. The implants showed the following survival rates after the artificial mouth: A2 and B2 100%; A3 and B3 87.5%. The following average fracture resistance values were found (fracture load [N]/bending moment [N mm]): A1 (583/2907), B1 (516/2825), A2 (618/2737), B2 (550/3150), A3 (802/3784) and B3 (722/3809). After 5 million loading cycles a significant increase in fracture load and bending moment was found. Modification of the abutment significantly decreased the fracture load of implants without foregoing dynamic loading. However, the shape modification altered the lever arm. For that reason, a smaller load resulted in the same bending moment. Therefore, abutment modification had no significant influence on the fracture resistance of ATZ.

Significance. Neither thermomechanical cycling in an aqueous environment nor modification of the abutment had a negative effect on the fracture resistance of ATZ.

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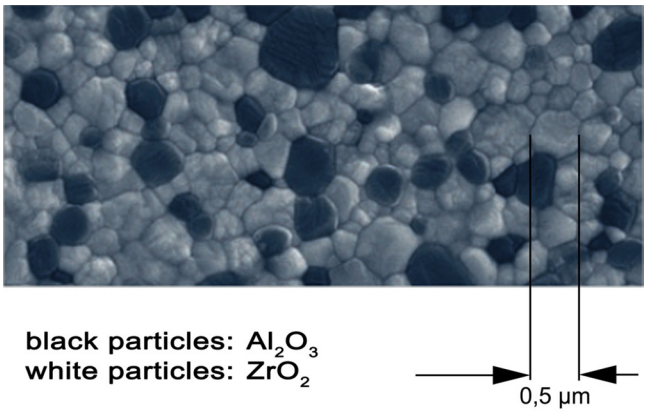

Fig. 1 – Structure of the ATZ (alumina-toughened zirconia) ceramic.

1. Introduction

Oral implants made of titanium have been shown to function well for many years [1–3]. However, the esthetic and biological properties of titanium have presented challenges and raised questions. The presence of thin peri-implant mucosa or soft-tissue recessions may result in visibility of the opaque-gray titanium implant. For the patient, this is particularly undesirable in the esthetically demanding anterior tooth region [4]. Furthermore, titanium residues have been detected in peri-implant soft-tissues and bone biopsies [5–9] and it has been postulated that titanium may lead to hypersensitization [10,11].

Newly developed high-performance ceramics of zirconium dioxide (ZrO_2) may be an alternative to titanium as an implant material. These ceramics possess good initial mechanical strength [12], exhibit favorable tissue compatibility [13–15], and show osseointegration comparable to that of titanium [16–24]. A further advantage of zirconia is the reduced plaque accumulation and greater resistance to mechanical processing [25,26]. Moreover, the white-opaque ZrO_2 ceramic resembles the tooth in terms of color, and thus provides good esthetics even with a thin gingiva or with soft-tissue recessions [27,28]. Zirconia has been successfully used in dentistry for restorations such as root posts [29,30], crowns and bridges [31–34], and implant abutments [35,36]. The use of zirconia in oral implantology and fixed implant prosthodontics [37] is still in its developmental stages and little research has been conducted so far for oral implants regarding the mechanical stability and ceramic aging [38,39]. The aging of zirconia can result in a decrease in the initially high flexural strength which may lead to fatigue fractures under "normal" masticatory loading [40]. Zirconia implants may undergo slow degradation during long term implantation in the human body [41]. A zirconium dioxide ceramic reinforced with alumina [ATZ (alumina-toughened zirconia, with 20 wt% alumina); Fig. 1 and Table 1] may circumvent these limitations [42]. Alumina-toughened zirconia has higher toughness values than yttria-stabilized zirconia. Additionally, increased alumina acts to constrain the zirconia particles, retaining the tetragonal zirconia in a metastable state, resulting in toughening the ceramic implant material. Moreover, the hardness of composites with increased alumina volume is greater, since alumina is harder than zirconia. This should lead to higher mechanical stability and lower aging of ATZ [38]. Finally, ATZ was shown to have no negative influence on behavior or differentiation of surrounding cells [43].

Table 1 – Material properties according to the manufacturer (ATZ: alumina-toughened zirconia; TZP: tetragonal zirconiumdioxide polycrystal).

Characteristics	Unit	TZP	ATZ
Components		ZrO_2/Y_2O_3	$ZrO_2/Al_2O_3/Y_2O_3$
Composition	wt%	95/5	76/20/4
Density	g/cm^3	6.05	5.5
Grain size	µm	0.4	0.4
Bending strength	MPa	1.000	2.000
Compressive strength	MPa	2.000	2.000
Young's modulus	GPa	200	220
Fracture toughness	$MPa\,m^{1/2}$	8	8

There is limited data on the stability of oral implants fabricated of ATZ [42]. The purpose of this investigation was to evaluate the thermomechanical stability of a commercially available ATZ implant system before and after artificial loading conditions in an aqueous environment over a simulated time period up to 20 years. Since one-piece implants often need intraoral final modification and given the fact that grinding is recognized to influence the strength of zirconia [44–47], an additional aim of this study was to investigate whether modification of the abutment would reduce the fracture resistance of this implant. The null hypothesis of this study was that neither thermomechanical cycling in an aqueous environment nor modification of the abutment will decrease the fracture resistance of the ATZ ceramic implant.

2. Materials and methods

A total of 48 one-piece alumina-toughened zirconia ceramic implants (4.4 mm diameter, 12 mm intraosseous length, 2.6 mm shoulder height, 6 mm length of the abutment; Metoxit, Thayngen, Switzerland) were used for the experiment (Fig. 2a). Material properties, as indicated by the manufacturing company, are shown in Table 1. The implants were divided into two groups (Table 2). Group A consisted of twenty-four unmodified implants. The abutments of group B implants were shape modified according to the preparation guidelines of a central incisor (Fig. 2b). The two groups were further divided into three subgroups: subgroup 1 – eight implants that were not subjected to artificial loading in the chewing machine; subgroup 2 – eight implants that were subjected to 1.2 million loading cycles in the chewing machine with thermocycling; subgroup 3 – eight implants that were subjected to 5 million loading cycles with thermocycling. The preparation of group B implants was performed with a Gentlepower Lux handpiece (Gentlepower Lux 25 LP, KaVo, Biberach, Germany) at a maximum speed of 140,000 rpm and water cooling from a triple-port spray system with 50 ml/min. The primary preparation was executed with diamond-coated instruments with a grain size of 120 µm (Figure 6878.314.012; Brasseler Komet, Lemgo, Germany). The finishing of the preparation was performed with diamond instruments of a grain size of 40 µm

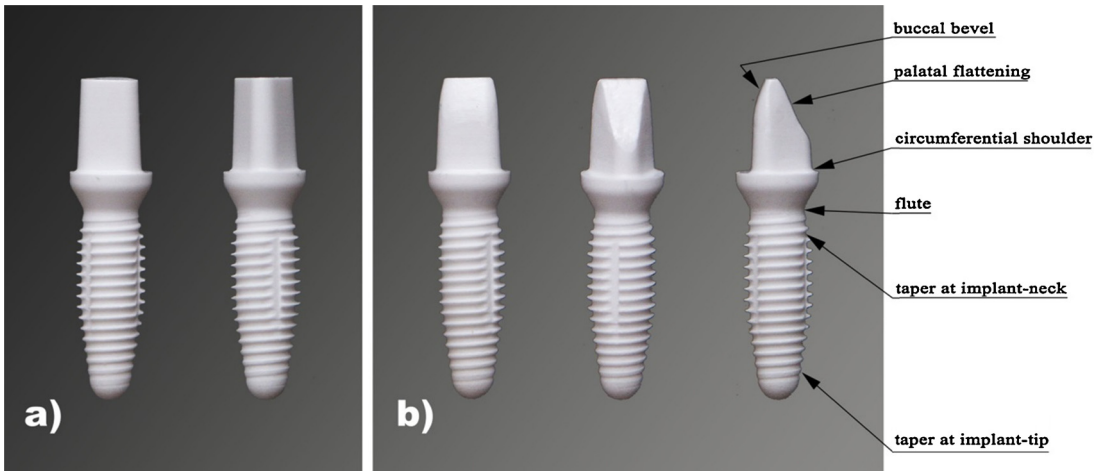

Fig. 2 – Implant without modification (a) and implant with modified abutment (b) according to the preparation guidelines of a central incisor.

Table 2 – Grouping of test and control specimens.

48 Ziraldent®-Implants					
Group A 24 Implants without modification			Group B 24 Implants with modified abutment		
A1 8 Implants 0 cycles	A2 8 Implants 1.2×10^6 cycles	A3 8 Implants 5×10^6 cycles	B1 8 Implants 0 cycles	B2 8 Implants 1.2×10^6 cycles	B3 8 Implants 5×10^6 cycles
↓	Dynamic loading test (chewing simulator)		↓	Dynamic loading test (chewing simulator)	
Static loading test					

(Figure 8878.314.012; Brasseler Komet). Since the implant-head consisted of a triangular rounded cone with a circumferential shoulder the modifications were limited to the preparation of a buccal bevel and a palatal flattening (Fig. 2b). A silicon index (Twinduo; Picodent, Wipperfuerth, Germany) of the master preparation served as control to ensure a standardized abutment shape for all modified implants.

The implants were embedded in an autopolymerizing acrylic resin (Technovit® 4000, Heraeus Kulzer, Wehrheim, Germany) in special sample holders. To realize a standardized procedure, two master sample holders were filled with a silicon material (Twinduo; Picodent, Wipperfuerth, Germany; Shore A hardness: 90). The use of an adjusted drilling jig allowed to drill a hole at an angle of 45° to the vertical, replicating the position of upper central incisors [48]. In order to represent a physiological clinical situation after one year (0.5–1 mm of bone remodeling [49,50]), two master implants (modified and unmodified) were screwed in the prepared hole up to the first implant thread. Silicon indices of these master models were used to embed 24 modified and 24 unmodified sample implants. The resin had a modulus of elasticity of approximately 12 GPa which approximates that of human bone (10–18 GPa) [51]. In order to compensate deviations due to the embedding procedure, standardized photographs in front of an adjusted grid were used to determine the individual embedding angle and lever arm for each of the 48 samples (Fraunhofer Institute for Mechanics of Materials, Freiburg, Germany; Fig. 3).

2.1. *Dynamic loading test*

Thirty-two of the specimens were thermomechanically aged in a computer-controlled dual axis-chewing simulator in an aqueous environment (Willytec, Munich, Germany; Fig. 4) in order to simulate five years (1.2 million cycles; subgroups A2, B2) and almost twenty years (5 million cycles; subgroups A3, B3) of clinical service, assuming an annual masticatory performance of 240,000–250,000 occlusal contacts [52]. The chewing simulator-environment consisted of eight identical sample chambers, two stepper motors controlling vertical and horizontal movements of the antagonists (Steatit® ceramic balls, 6 mm in diameter, Hoechst Ceram Tec, Wunsiedel, Germany) against the implant samples, and a hot and cold water circulation system (Haake, Karlsruhe, Germany). The antagonist ball had a Vickers hardness similar to that of enamel [53]. The applied load in the chewing simulator was 98 N (10 kg) [54,55] and the point of load application on the implants was placed on the palatal upper edge. This resulted in an elongated lever arm for the modified samples. The load was applied onto the implants by combined vertical (6 mm) and horizontal (0.5 mm) movements, which – via computerized adaptation – represented an approximation to the physiological masticatory cycle of axial pressure and horizontal shear. The cyclic loading was set at 1.6 Hz. The thermocycling was from 5 °C to 55 °C for 60 s each with an intermediate pause of 12 s, maintained by the thermostatically-controlled liquid circulator (Haake, Karlsruhe, Germany). During the dynamic loading,

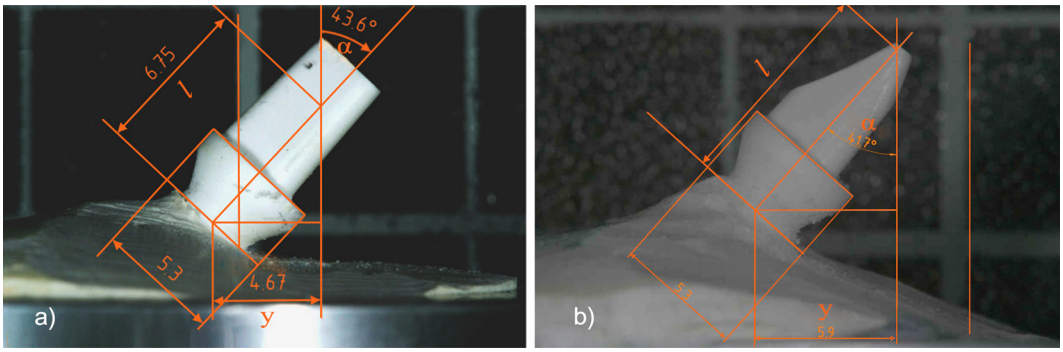

Fig. 3 – Standardized photographs of the embedded implants without (a) and with (b) abutment modification allowed the calculation of the lever arm ($y = \sin\alpha \cdot l$).

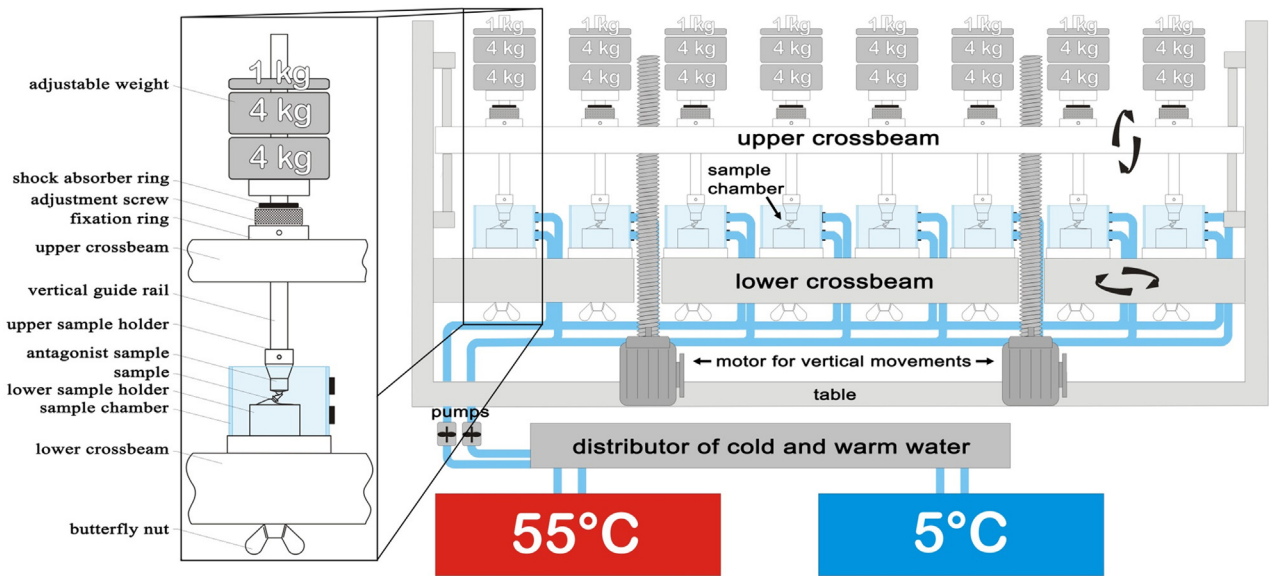

Fig. 4 – Schematic drawing of the chewing simulator (Willytec, Munich, Germany). The vertical guide rail and the sample holder weigh another 1 kg.

Table 3 – Settings of the chewing simulator machine.

Chewing cycles	1,200,000/5,000,000
Cycle frequency	1.6 Hz
Vertical movement	6 mm
Horizontal movement	0.5 mm
Descending speed	60 mm/s
Rising speed	55 mm/s
Forward speed	60 mm/s
Backward speed	55 mm/s
Applied weight per sample	10 kg (98 N)
Hot dwell time	60 s
Hot bath temperature	55 °C
Cold dwell time	60 s
Cold bath temperature	5 °C
Intermediate pause	12 s

all samples were examined twice a day. The chewing machine needed approximately 9 and 36 days to accomplish 1.2 million cycles (Group B) and 5 million cycles (Group C). Fractures of the implants were recorded as a failure. The details of the settings of the chewing simulator machine are listed in Table 3.

2.2. *Static loading test*

All samples that survived the exposure to the chewing simulator without fracture were statically loaded to fracture using a universal-testing machine (Zwick, Z010/TN2S, Ulm, Germany). All samples were loaded using the same sample holders and at the same contact point as used for the dynamic loading. A vertical compressive load was applied on the palatal side of the angulated implants under a crosshead speed of 10 mm/min. The loads required for fracturing the samples were recorded using the X-Y writer of the Zwick testXpert® V 7.1 software, with failure recorded at the first sharp drop-down of the graphical curve (fracture of the ceramic, Fig. 5). A graph was drawn for each sample representing and was used to determine the load at time of fracture.

2.3. *Statistical analysis*

Mean values of the acquired data (fracture load, lever arm and bending moment) of each subgroup (A1, A2, A3 and

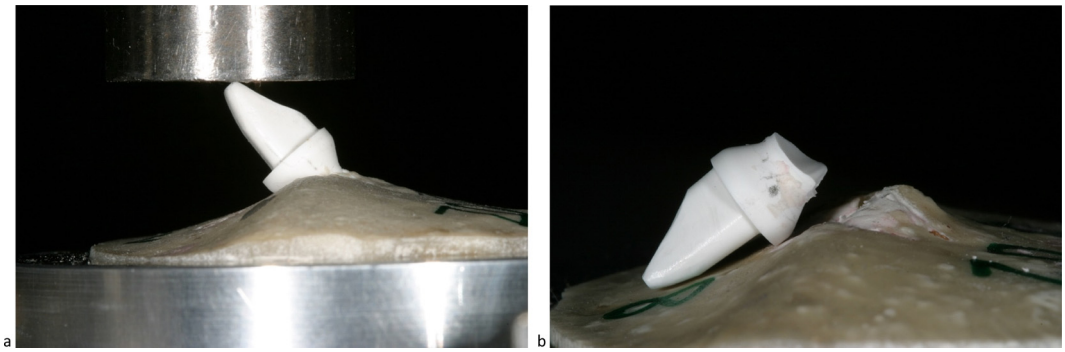

Fig. 5 – Static loading test of an implant with modified abutment (a); site of fracture after loading (b).

B1, B2, B3) were calculated to create graphs of the resulting confidence intervals. A two-way analysis of variance (ANOVA) was used. The continuous response variables (fracture load, lever arm and bending moment) were modeled as a function of abutment modification (w = with modification; w/o = without modification), cycles (0, 1.2 million cycles, 5 million cycles) and the corresponding interaction as explanatory variables. Pairwise differences of least-square means were calculated and *p*-values were adjusted by the method of Tukey. The level of significance was set at $p < 0.05$. The Wilcoxon-Test allowed multiple pairwise comparisons of different subgroups regarding their response variables. All computations were performed with the statistical software SAS (SAS system v9.1; SAS Institute Inc., Cary, NC).

3. Results

3.1. *Dynamic loading test*

All implants of the subgroups A2 and B2 (1.2 million loading cycles) survived the dynamic loading test resulting in 100% survival. One implant from group A3 and group B3 failed over the course of the dynamic loading test. The failure of the implant in group A3 occurred after 275,000 cycles, while the failure of the implant in group B3 failed after 250 cycles. Thus, there was an 87.5% survival rate in both groups with 5 million loading cycles.

3.2. *Static loading test*

All implants fractured in a pattern similar to that shown in Fig. 5b. The values for lever arm, fracture load and bending moment are given in Table 4. For the two implants which did not survive the dynamic loading test fracture load values were set to 98 N for statistical analysis due to their failure in the chewing simulator while being loaded with 10 kg. The calculated values of the fracture load, the bending moment and the lever arms were illustrated in the form of 95% confidence intervals (Figs. 6–8).

In the two-way analysis of variance, the fracture load and bending moment values showed statistically significant differences when comparing the different dynamic loading protocols (0, 1.2, 5 million cycles; $p < 0.006$). Pairwise comparison of the post hoc analysis showed statistical significance between 0 and 5 million cycles as well as between 1.2 and 5 million cycles. No significance between 0 and 1.2 million cycles was found. Modification of the abutment had no significant influence on fracture load ($p = 0.1540$) and bending moment ($p = 0.6352$). As well, interaction of modification and dynamic loading protocol had no significant influence on fracture load ($p = 0.9926$) and bending moment ($p = 0.6931$). Regarding the lever arm, there were no statistically significant differences when comparing the different dynamic loading protocols (0, 1.2, 5 million cycles; $p = 0.6361$). Modification of the abutment significantly altered the lever arm ($p < 0.001$), whereas the interaction of modification and dynamic loading

Table 4 – Means of measurements (embedding angle, lever arm, fracture load and bending moment) for the different test groups ($M = y \cdot F$; $y = \sin\alpha \cdot l$) (w = with, w/o = without, *n* = number of specimens).

Cycles	Modification	Subgroup	*n*	Embedding angle, α		Lever arm, y [mm]		Fracture load, *F* [N]		Bending moment, *M* [N mm]	
				Mean	SD	Mean	SD	Mean	SD	Mean	SD
0	w/o	A1	8	44.1	2.1	5.0	0.3	583.1	101.4	2907.3	532.0
	w	B1	8	38.6	2.0	5.5	0.3	515.8	35.6	2824.8	262.2
1.2×10^6	w/o	A2	8	41.5	7.0	4.5	0.9	618.2	83.0	2737.4	492.2
	w	B2	8	41.3	5.4	5.8	0.8	549.6	53.9	3150.2	347.6
5×10^6	w/o	A3	8	44.0	1.4	4.7	0.2	802.4	298.1	3784.0	1373.6
	w	B3	8	37.8	1.4	5.4	0.3	721.7	260.4	3808.5	1344.5

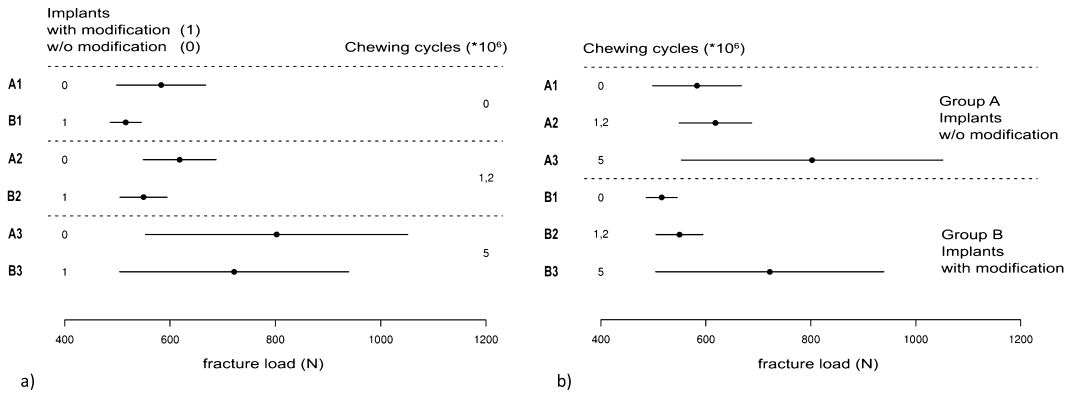

Fig. 6 – 95% confidence intervals of the fracture load ordered by abutment configuration (a) and group affiliation (b).

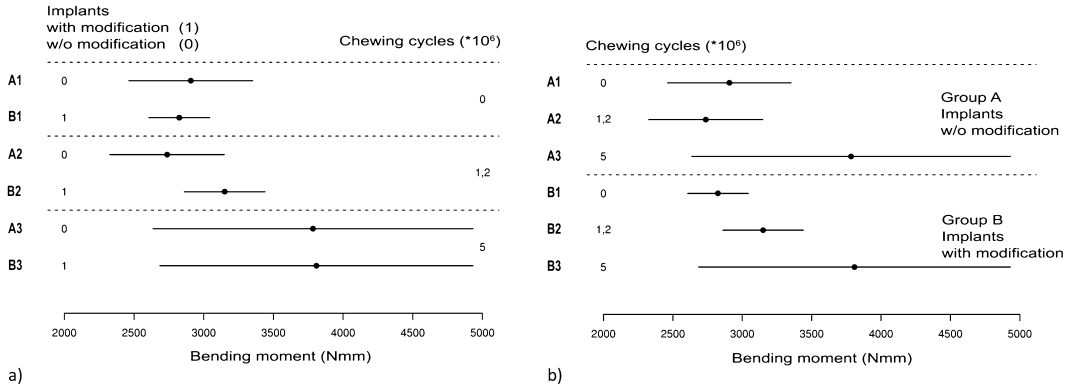

Fig. 7 – 95% confidence intervals of the bending moment ordered by abutment configuration (a) and group affiliation (b).

protocol had no significant influence on the lever arm ($p = 0.0964$).

The results of the Wilcoxon-Tests are listed in Table 5. Four pairwise comparisons of selected subgroups regarding their response variables (fracture load, bending moment and lever arm) have been performed:

(1) Influence of modification without any dynamic loading (A1:B1).
(2) Influence of modification after 5 million loading cycles (A3:B3).
(3) Influence of dynamic loading on implants without modification (A1:A3).
(4) Influence of dynamic loading on implants with modification (B1:B3).

Modification of the abutment significantly reduced the fracture load without forgoing dynamic loading. However, this observation was not applicable after 5 million loading cycles. There was no significant difference between prepared and unprepared implants regarding the bending moment. Modification of the abutment significantly increased the length of the lever arm before and after dynamic loading. It could

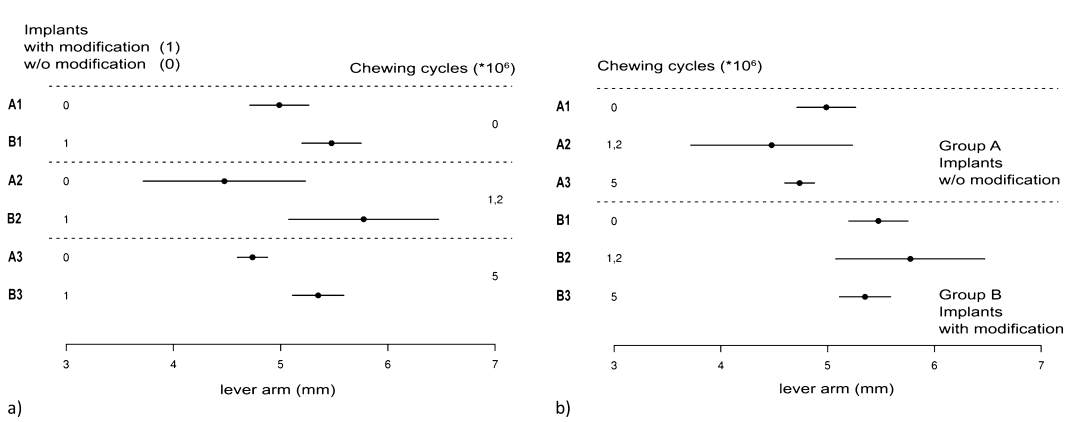

Fig. 8 – 95% confidence intervals of the lever arm ordered by abutment configuration (a) and group affiliation (b).

Table 5 – Pairwise comparisons of selected subgroups regarding their response variables (fracture load, bending moment and lever arm).

Response variable	Comparison	*p*-Value	Significance
Fracture load	A1:B1	0.0321	Significant
	A3:B3	0.2265	Not significant
	A1:A3	0.0261	Significant
	B1:B3	0.0261	Significant
Bending moment	A1:B1	0.1486	Not significant
	A3:B3	0.9588	Not significant
	A1:A3	0.0261	Significant
	B1:B3	0.0261	Significant
Lever arm	A1:B1	0.0255	Significant
	A3:B3	0.0052	Significant
	A1:A3	0.0627	Not significant
	B1:B3	0.4409	Not significant

be revealed that dynamic loading increased the fracture load and the bending moment significantly among the prepared implants as well as the unprepared implants. The dynamic loading protocol had no significant influence on the length of the lever arm.

4. Discussion

Currently, there is laboratory data available on the biomechanical characteristics of a "prototype" ceramic implant fabricated out of ATZ [42]. The purpose of the present study was to examine the fracture stability of a commercially available ATZ one-piece implant (Ziraldent® FR1; Metoxit, Thayngen, Switzerland). The effects of abutment modification and different thermomechanical cyclic loadings in an aqueous environment were evaluated.

Masticatory simulation trials represent an important method for pre-clinical study of materials and devices, and have been used for the study of several all-ceramic restorations [42,56–58]. They should imitate the occlusal loading, create forces comparable to those which develop during horizontal and vertical components of masticatory motion and re-create environmental factors such as temperature and moisture fluctuations as found in the oral cavity [59]. To simulate these conditions to the extent possible, the experimental setup of the current investigation differed from ISO 14801, which does not include horizontal components of a dynamic loading or environmental factors as mentioned before. Furthermore, ISO 14801 dictates the simulation of a 3 mm bony recession and an angle of 30° to the vertical. The implants of the current testing were embedded with an angle of 45° to the vertical and up to the first implant thread, simulating 0.5–1 mm of bony recession. Therefore, the calculated fracture load values are not comparable to other investigations adapting ISO 14801. Nevertheless, the calculation of the bending moment includes both mentioned variables and is, therefore, more crucial and the significant value when comparing different investigations, respectively.

Modified implants not exposed to dynamic loading showed significantly decreased fracture load values. This difference was not noted after 5 million cycles of dynamic loading. However, considering only the fracture load ignores the influence of lever arm extension. Therefore, lever arms of every sample have been measured to allow the calculation of the bending moment to fracture at the implant neck area (Fig. 9). The modified implants had lever arms which were significantly longer resulting in an increased leverage effect. In the present of a steady load, a longer lever arm will cause an increased bending moment at the implant neck area leading to a higher destructive force. Thus, for the modified implants, smaller forces resulted in the same bending moment as for the unmodified implants. When solely considering the fracture load values of this investigation, it seemed that modification of the abutment decreased the stability of the ATZ ceramic. However, incorporation of the individual lever arm revealed no significant difference of the bending moment at the timepoint of ceramic fracture between modified and unmodified samples. In summary, the stability of the ATZ implants is not reduced through modification, but the modified shape of the abutment led to a longer lever arm when the bending force was introduced. This reduced the amount of force that led to fracture.

In this study, the load applied to all implants during masticatory simulation was 98 N. This force was chosen to simulate the physiological loading of maxillary teeth, following a clinical investigation by Fontijn-Tekamp and coworkers who found normal chewing forces of 60–75 N in the anterior dentition, and 110–125 N in the posterior dentition [54]. Two implants fractured during masticatory simulation under 98 N of dynamic loading (A3: 250 cycles, B3: 275.000 cycles). A failure in the production process of these ceramic implants resulting in distinct flaws could be an explanation for the early fracture. The associated lever arms (A3: 4.8 mm; B3: 5.9 mm) showed no noticeable deviation from the ones of the other samples. Regrettably, it might be a coincidence that these samples failed in the early stage of the long-term testing (subgroup 3) and, therefore, the presented survival rates might be misleading. Due to their failure long before 1.2 million cycles the two mentioned samples could have equally been assigned to subgroup 2 (short-term testing) or even subgroup 1 (static loading) and, therefore, fail under minimal load in the universal testing machine. Since the sample assignment to the different subgroups was random and before any testing procedure, this interrogation cannot be clarified afterwards. The early failure of the two implants in subgroup 3 is certainly well outside the values found for fracture load of all other implants in the study. These two "outliers" reduced the mean values and increased the variability (SD) in the fracture resistance for these two groups. Still the results of subgroup 3 showed significantly increased mean values of fracture load and bending moment. Therefore, it can be assumed that the results of the static loading test and their interpretations remain unaffected by these considerations and, therefore, are of major importance then the presented survival rates of the dynamic loading test.

As it can be seen in Table 4, the variability in the length of the lever arm of subgroup 2 (A2, B2) was substantially larger than in subgroups 1 and 3. This was due to one sample in each group which showed major deviations of the embedding angle (A2/Sample 10: 28°, B2/Sample 6: 52°), resulting in a shortened (Sample 10: 2.9 mm) and elongated (Sample 6: 7.4 mm) lever arm, respectively. This reveals the susceptibility for inaccuracies of the applied method for a standardized embedding procedure using silicon indices of a master

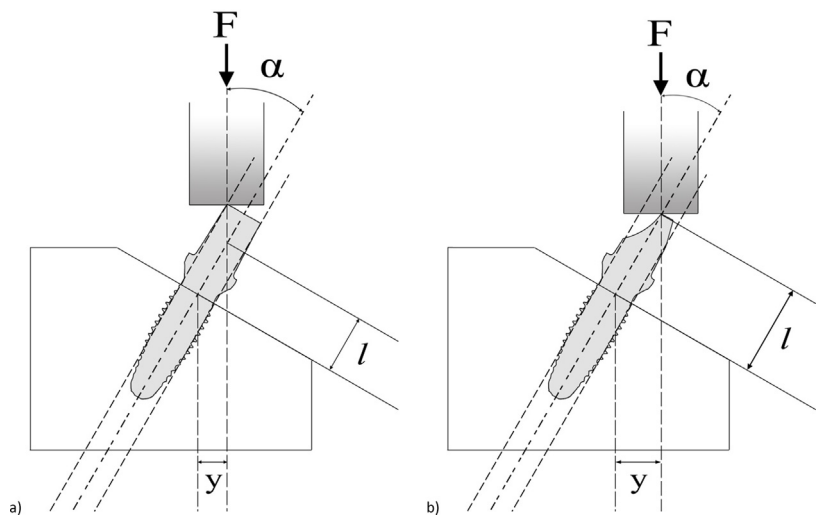

Fig. 9 – Calculation of the lever arm (y) and the bending moment (M) at the timepoint of implant fracture in the static loading test ($M = y \cdot F$; $y = \sin\alpha \cdot l$). Abutment individualization (b) resulted in a longer lever arm compared with unmodified samples (a).

sample. This method has been improved for further research. Nevertheless, these deviations of the embedding angle had no significant influence on the average fracture load (Sample 10: 574 N, Sample 6: 502 N).

The mean fracture load values for the implants of all subgroups (516–812 N) exceeded the maximum chewing forces. In the posterior dentition, maximum chewing forces from 250 to 400 N have been measured, and in the anterior dentition they ranged from 140 to 170 N [54]. According to Schwickerath [60], the initial strength of an anterior restoration should exceed 400 N to withstand recurrent loadings in that region. He considered potential aging processes caused by undercritical crack propagation when drafting this requirement. Körber and Ludwig [61] claimed a further safety margin of at least 200 N added to the maximally chewing forces of 200 N for anterior and 300 N for posterior teeth, respectively. According to the mentioned postulations for ceramic restorations, the results of the current investigation should be adequate for a clinical application. Finally, the mean bending moment to the timepoint of fracture in the current investigation exceeded values that are known to occur at implant sites intraorally to the factor 10 or more [62,63].

Kohal et al. [42] evaluated the survival and fracture load of an ATZ-implant prototype design with and without modification. The implants showed a statistically significant decrease in fracture load after 5 million cycles of artificial loading compared with non-loaded implants. The mean fracture load dropped from 1734 N to 1358 N. In contrast to the evaluation of the prototype design, the 20-year masticatory simulation of the current investigation resulted in a significantly increased fracture load for modified and unmodified implants. Due to the metastability of tetragonal zirconia, stress-generating procedures like cyclic loading in an aqueous environment are able to trigger the t → m transformation with the associated volume increase leading to the formation of surface compressive stresses, thereby increasing the flexural strength but decreasing the resistance to aging [64,65]. Besides this phase transformation toughening mechanism, preexisting different fracture load values of the Ziraldent®-Implants or altered viscoelastic and/or mechanical properties of the artificial bone due to the thermomechanical cycling in an aqueous environment might be further possible reasons for the results of the current investigation. It cannot be ruled out that group A3 and B3 implants already featured higher fracture resistance prior to their dynamic loading trial. Nevertheless, the significant increase in fracture resistance of this implant system after long-term thermomechanical cycling in an aqueous environment could be confirmed in a separate verification (unpublished data) with an improved embedding procedure and an alternate embedding resin. Further investigations and profound methodology (e.g. Raman spectroscopy and/or X-ray diffraction) is needed to clarify this phenomenon. Throughout, fracture load values measured in the current investigation are inferior to those of the previously tested prototype design from ATZ. This difference in fracture load might have several reasons: (1) the implant design has changed including a reduced diameter at the thinnest area (i.e. the implant neck) from 3.7 to 3.5 mm, (2) the prototype surface was left as machined whereas the final implant underwent several surface modification procedures. Since implant fracture resistance is directly correlated with the third power of the implant diameter, a diameter reduction of 0.2 mm can account for an approximately 15% diminished fracture resistance. Features of the implant design like the configuration of the threads can also contribute to its fracture resistance [66]. Sharp threads as well as internal line angles at the junction of the thread with the implant body allow for focal areas of high stress concentration leading to fracture [67]. However, there have been no changes in the geometry of the prototype's thread design. Surface treatments may compromise the shear and tensile strength of ceramic implants and have, therefore, as well been suggested as potential causes for a reduced fracture resistance. [68]. The prototype's surface was left as machined whereas the implant of the current investigation underwent surface modifications to improve osseointegration in vivo. Surface modifications of zirconia oral implants are generally compromised of an

initial sandblasting procedure in terms of a "priming" followed by the application of ceramic slurry that produces a porous surface after a final sintering process. Sandblasting as well as the additional sintering process may contribute to the reduced fracture resistance of the final implant design [69].

5. Conclusions

All ATZ implants in the present investigation revealed sufficient values in the static fracture load test to withstand physiological bite forces in the oral cavity. Differences in fracture load were related to different lever arm lengths and not to modification of implants. Individual length of the lever arms results in variable leverage effects. Therefore, the calculation of the bending moment for each sample at the timepoint of fracture in the static loading test seems to be more valid compared to fracture load values alone. It could be shown, that neither 20-years of prior masticatory simulation in an aqueous environment nor abutment modification had a negative effect on the bending moment. Hence, the null hypothesis could not be rejected. Regarding their fracture resistance, the evaluated Ziraldent®-Implants can be recommended for clinical testing.

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