

Mechanical Performance of Cantilever Implant-Supported Fixed Partial Dentures Fabricated by Multilayer Zirconia

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Abstract

Objectives

To investigate the mechanical performance, particularly fracture strength, of implant-supported cantilevered fixed partial dentures (cFPDs) fabricated from multilayer zirconia compared with monolithic zirconia, and to evaluate the influence of connector dimensions on their mechanical performance.

Material and Methods

Forty cFPDs were digitally designed, milled, and divided into 4 groups ($n = 10$) according to zirconia type and connector dimension: 4×4 mm monolithic, 3×3 mm monolithic, 4×4 mm multilayer, and 3×3 mm multilayer zirconia. Specimens underwent thermocycling (5000 cycles, 5°C–55°C) followed by static loading until catastrophic fracture. The maximum load-to-fracture was recorded. Data were analyzed using 2-way ANOVA followed by Tukey post hoc tests ($\alpha = .05$).

Results

Both zirconia type and connector dimension influenced fracture strength. The 3×3 mm multilayer group had the lowest (109.3 ± 33.3 N), while 4×4 mm monolithic group showed the highest mean fracture load (464.3 ± 240.1 N). Failures occurred exclusively by fracture of zirconia framework, without damage to abutment screws or implant analogs. Most cracks occurred at the distal abutment wall and propagated toward the cantilever connector.

Conclusions

Connector dimensions and material composition critically affect cFPD fracture resistance. Multilayer zirconia restorations demonstrated lower fracture strength than monolithic zirconia, emphasizing the need for careful material selection and design optimization to enhance the durability of implant-supported cantilever prostheses.

Clinical Relevance:

The current findings suggest that clinicians should carefully consider zirconia type and connector dimensions when designing implant-supported cantilever restorations to reduce the risk of framework fracture and improve clinical longevity.

INTRODUCTION

Implant-supported cantilevered fixed partial dentures (cFPDs) have emerged as a widely accepted treatment option for patients with distal extension edentulous ridges, due to their advantages such as improved functionality, cost-effectiveness, and high patient satisfaction [1–4]. However, successful outcomes and longevity of Implant-supported cFPDs depend on multiple factors, including adherence to occlusal and biological principles, meticulous consideration of cantilever extension, and precise technical execution procedures. Paramount to this success is the selection of structural materials capable of withstanding occlusal forces while providing a stable and durable foundation for the restoration [5]. Traditionally, porcelain-fused-to-metal (PFM) restorations have seen the gold standard, offering excellent mechanical properties, acceptable aesthetics, and proven clinical durability [6–9]. Despite these advantages, PFM restorations are associated with several drawbacks, including technical complexity, susceptibility to chipping of the veneering material, and time-consuming fabrication processes. These limitations have contributed to a shift toward modern materials and digital workflows [10, 11].

Among contemporary materials, zirconia, a high-performance ceramic, has gained increasing popularity in recent years due to advancements in computer-aided design and manufacturing (CAD-CAM) technologies. Zirconia is well-known for its exceptional biocompatibility, high fracture strength, reliable reproducibility, low bacterial adhesion, and aesthetic properties, making it an ideal material for prosthetic frameworks, including cFPDs [12–14]. Also, reproducibility is beneficial if any complication arises, due to the advantages of the CAD- CAM workflow. Mechanically, zirconia exhibits impressive mechanical properties, with fracture toughness between 6 to 10 MPa√m and flexural strength between 900 and 1200 MPa [15–17]. Dental zirconia is available in various formulations, primarily differentiated by its yttria content (Y2O3), which significantly influences its phase composition, mechanical properties, and optical characteristics. These formulations include 3Y-TZP (3 mol% Y2O3), 4Y-TZP (4 mol% Y2O3), 5Y-TZP (5 mol% Y2O3), and 6Y-TZP (6 mol% Y2O3).[18–21] Monolithic 3Y-TZP (3 mol% Y2O3) is predominantly tetragonal and exhibits excellent strength, making it ideal for high-stress posterior restorations, but its limited translucency can be a disadvantage for anterior aesthetics. Increasing yttria content produces partially stabilized zirconia (PSZ), such as 4Y-PSZ and 5Y-PSZ, which combine tetragonal and cubic phases to improve translucency while moderately reducing strength [22]. Multilayer zirconia incorporates these hybrid compositions in polychromatic layers, aiming to mimic natural tooth aesthetics without veneering porcelain [18, 21].

Connector dimensions also play a critical role in the fracture resistance of zirconia cFPDs, with larger connectors reducing localized stress and improving structural durability [12, 14]. Although previous studies have investigated the fracture resistance of monolithic zirconia cFPDs, many focused on single zirconia formulations or uniform connector designs. Consequently, comparative data on multilayer zirconia with varying connector dimensions remain limited. A clearer understanding of the mechanical behavior of multilayer zirconia is therefore essential to guide material selection and prosthesis design.

Therefore, this study seeks to address this knowledge gap by evaluating the effect of zirconia type and connector size to the fracture strength of cFPDs. Multilayer zirconia materials with hybrid compositions

(4Y-PSZ and 5Y-PSZ) were compared with monolithic zirconia of uniform composition (3Y-TZP). Two connector cross-sectional dimensions were investigated: 4 × 4 mm (16 mm²) and 3 × 3 mm (9 mm²). The null hypotheses tested in this study were that there is no significant difference in fracture strength between multilayer and monolithic zirconia cFPDs, and that connector cross-sectional dimension does not significantly affect fracture strength.

MATERIALS AND METHODS

The mechanical behavior of implant-supported cFPDs was assessed with respect to zirconia composition and connector cross-sectional dimensions. A power analysis (G*Power; Ver. 3.1.9.7; Heinrich-Heine-Universität Düsseldorf) determined the sample size at a power of 0.8 at the significance level of 0.05. A total of 40 zirconia implant-supported cFPD with a 9mm distal cantilevers, were divided into 4 test groups (n = 10) based on connector dimension (4×4 mm/16 mm² and 3×3 mm/9 mm²) and zirconia type: monolithic 3Y-TZP (IPS e.max ZirCAD LT; Ivoclar Inc.) and multilayer 4Y-PSZ and 5Y-PSZ (ZirCAD Prime Esthetic; Ivoclar Inc.) zirconia (Table 1).

Table 1
Materials used for each group.

Groups	Material	Manufacturer	Composition	Connector Dimension
LT 4×4	Zirconia; IPS e.max ZirCAD LT	Ivoclar Inc.	zirconium oxide raw materials (3Y-TZP)	4×4 mm/16 mm ² (n = 10)
LT 3×3	Zirconia; IPS e.max ZirCAD LT	Ivoclar Inc.	zirconium oxide raw materials (3Y-TZP)	3×3 mm/9 mm ² (n = 10)
PE 4×4	Zirconia; IPS e.max ZirCAD Prime Esthetic PE	Ivoclar Inc.	Blend of 2 zirconium oxide raw materials (4Y-TZP and 5Y-TZP)	4×4 mm/16 mm ² (n = 10)
PE 3×3	Zirconia; IPS e.max ZirCAD Prime Esthetic PE	Ivoclar Inc.	Blend of 2 zirconium oxide raw materials (4Y-TZP and 5Y-TZP)	3×3 mm/9 mm ² (n = 10)
LT, IPS e.max ZirCAD LT; PE, ZirCAD Prime Esthetic.				

A three-dimensional (3D) printed resin block model (Gray Pro resin; Formlabs), measuring 35×20×16.3 mm (Fig. 1A), was digitally designed using Autodesk Inventor software to house 2 implant analogs with 11 mm length (Bone Level RC, TAN; Straumann), representing the sites of the right mandibular first premolar and first molar. The analogs were positioned 15mm apart (center to center) and cemented into the model block using resin cement (Multilink hybrid abutment; Ivoclar Inc.) under constant pressure of 200 grams, following the manufacturer’s instructions. A 2° mesiodistal angulation was applied to the

premolar analog, while the molar analog remained parallel to the long axis, simulating typical clinical limitations in posterior mandibular implant placement.

Scan bodies (Elos Accurate Intra Oral Scan Body 4.1/4.8 RC; Elos Medtech) were attached to the implant analogs (Fig. 1B) and scanned by a Lab scanner (Medit T-Series; MEDIT). Digital models of implant-supported cFPDs were created using CAD software (Dental System; 3Shape), incorporating 2 connector dimensions: 4×4 mm/16 mm² and 3×3 mm/9 mm². For the first premolar site, an engaging abutment (Ti Base Engaging, 4.1/4.8 RC, 2 mm GH; DESS) was selected, while a non-engaging abutment (Ti Base Non-Engaging, 4.1/4.8 RC, 2 mm GH; DESS) was used for the first molar. Then 2 STL files were generated for each design, differing only in connector dimension.

Each connector design was fabricated using 2 types of zirconia materials: 1) LT: monolithic 3Y-TZP zirconia (IPS e.max ZirCAD LT; Ivoclar Inc.) and 2) PE: multilayer 4Y-PSZ +5Y-PSZ zirconia (ZirCAD Prime Esthetic; Ivoclar Inc.), with 10 specimens for each material. Milling was performed using a 5-axis machine (PrograMill PM7; Ivoclar Inc.) with nesting using PrograMill CAM software. For multilayer zirconia, connectors were positioned within the 4Y dentin zone. Sintering followed manufacturer protocols (Programat S1 1600; Ivoclar Inc.). Prior to assembly, titanium bases and zirconia intaglio surfaces were air-abraded with 50 µm Al₂O₃ at 2 bar pressure (Basic Quattro IS, Renfert Inc.) from a 10 mm distance for 20 seconds. All components were cleaned in ethanol using an ultrasonic bath for 10 minutes and air-dried. A universal primer (Monobond Plus; Ivoclar Inc.) was applied to both bonding surfaces per manufacturer guidelines. Titanium bases were torqued to 35 Ncm onto their respective implant analogs (Fig. 1C), with PTFE tape sealing the screw channels. The zirconia restorations were then luted to the titanium bases using resin cement (Multilink Hybrid Abutment; Ivoclar Inc.) under 200 g of constant pressure, and excess cement was removed.

Specimens were stored in distilled water at 37°C for 24 hours, then thermal aged (Huber; SD Mechatronik) for 5 000 cycles between 5°C and 55°C, with 30-second dwell and 15-second transfer times. Post-thermocycling, specimens were mounted on a universal testing machine (5567; Instron Ltd) and secured with clamps. Occlusal contact points were identified using articulating paper (AccuFilm; Parkell), and static vertical loading was applied using a 9 mm stainless-steel ball placed 2 mm distal to the distal marginal ridge of the pontic. A 2 000-N load cell was used for the load-to-failure test, applying force at a crosshead speed of 2 mm/min at 90° to the occlusal plane (Fig. 1D). Maximum fracture force (N) was recorded at the point of a sudden force drop. Failure was confirmed visually. The fractured surfaces were examined using the stereo microscope (Olympus SZX12; Olympus Microscopes) and scanning electron microscopy (SEM) (Hitachi SU70; HITACHI).

Prism software (Version 9; GraphPad) was used to perform statistical analysis at a significance level of 0.05. Data normality was assessed via the Kolmogorov-Smirnov test. A 2-way ANOVA (Prism Version 10; GraphPad Software) was conducted to evaluate the effects of connector dimension and zirconia type, followed by Tukey post hoc analysis.

RESULTS

The Kolmogorov-Smirnov normality test confirmed that the data followed a normal distribution ($P > .05$). The 2-way ANOVA test demonstrated a statistically significant difference, indicating that both the zirconia type and connector dimension influence the maximum load of force ($P < .05$), with no significant interaction between 2 factors (Table 2). The lowest value of fracture strength was found in the PE 3×3 mm ($109.3 \pm 33.33\text{N}$), which had the smaller connector dimension and multilayer (4Y-PSZ + 5Y-PSZ) zirconia material. Group LT 4×4 mm, with monolithic 3Y-TZP zirconia and increased connector cross-sectional dimension, demonstrated the highest values of fracture strength $464.3 \pm 240.1\text{ N}$ (Fig. 2). The fracture strength values were significantly decreased by material type (4Y-PSZ + 5Y-PSZ < 3Y-TZP) and decreased connector dimensions (3×3 mm < 4×4 mm) (Fig. 2).

Table 2
The analysis results of 2-way ANOVA

Interaction	SS	DF	MS	F (DFn, DFd)	P value
	308	1	308	F (1,36) = 0.01	0.91
Zirconia Type	279726	1	279726	F (1,36) = 12.64	0.001
Connector Dimension	352501	1	352501	F (1,36) = 15.93	P < 0.001
Residual	796747	36	22132		

All specimens fractured without any damage or plastic deformation of the abutment screw or analog. Table 3 summarizes the distribution of fractures within each group. Most of the zirconia specimens exhibited fractures at the distal abutment wall, with crack propagation progressing catastrophically along this wall toward the cantilever connector region (Fig. 3A). Figure 3B and Fig. 3C show the crack pattern observed under the stereo microscope. Figure 3D-F revealed the presence of compression curls in the connector region, showing the cracks extended transversely through the thickness of the distal abutment wall, ultimately reaching the connector region.

Table 3
Distribution of fractures within each group

Zirconia Type	Connector Dimension	At the distal abutment	At the mesial abutment
LT	4×4	n = 9	n = 1
	3×3	n = 7	n = 3
PE	4×4	n = 8	n = 2
	3×3	n = 7	n = 3
LT, IPS e.max ZirCAD LT; PE, ZirCAD Prime Esthetic.			

DISCUSSION

The null hypotheses were rejected, as both zirconia composition and connector cross-sectional dimension significantly influenced the fracture strength of implant-supported cantilevered fixed partial dentures (cFPDs). Monolithic 3Y-TZP zirconia demonstrated significantly greater fracture resistance than multilayer zirconia with a hybrid composition (4Y-PSZ + 5Y-PSZ). In addition, restorations with larger connector dimensions (4×4 mm/16 mm²) exhibited significantly greater fracture strength than those with smaller connector dimensions (3×3 mm/9 mm²). These findings emphasize the critical influence of both material composition and connector geometry on the mechanical performance of implant-supported cFPDs.

The fracture resistance values obtained in the present study highlight the combined effect of zirconia composition and connector dimension on the structural durability of implant-supported cFPDs. Monolithic 3Y-TZP zirconia with 4×4 mm connectors demonstrated the highest fracture resistance, whereas multilayer zirconia with 3×3 mm connectors demonstrated the lowest fracture resistance. Reported maximum occlusal forces in the posterior region generally range between 300 and 700 N, while cantilevered prostheses are subjected to increased bending moments and stress concentration during function [23]. Although some specimens fractured below the maximum occlusal force range reported for cantilevered prostheses [23], restorations fabricated from monolithic zirconia with larger connector dimensions exceeded the average maximum bite force reported in non-bruxing individuals.[24] Nevertheless, these values may remain insufficient under severe parafunctional loading conditions, such as bruxism, where occlusal forces can substantially exceed physiologic levels and may reach up to 847 N in men and 597 N in women in the molar region [25]. Consequently, careful consideration of zirconia composition and connector dimensions is essential when designing implant-supported cFPDs for patients with increased occlusal demands, particularly in posterior regions and in individuals with parafunctional habits.

The superior mechanical performance observed in monolithic 3Y-TZP zirconia may be attributed to its predominantly tetragonal crystalline phase and associated transformation toughening mechanism. During crack propagation, stress-induced transformation from the tetragonal to monoclinic phase results in localized volumetric expansion, which impedes crack propagation and enhances fracture toughness [26, 27]. In contrast, multilayer zirconia materials with increased yttria content contain a greater proportion of cubic phase particles. Although this phase composition improves translucency and esthetic properties, it reduces transformation toughening capacity and consequently compromises mechanical strength [18]. The lower fracture resistance observed in the multilayer zirconia groups is therefore consistent with previous studies demonstrating the inverse relationship between translucency and fracture resistance in zirconia-based ceramics [28].

Connector cross-sectional dimension also demonstrated a substantial effect on fracture resistance. Specimens fabricated with 4×4 mm connectors consistently exhibited significantly greater fracture strength than specimens with 3×3 mm connectors. These findings are consistent with previous

investigations reporting that increased connector dimensions reduce stress concentration and improve structural integrity in zirconia restorations [12, 14]. The increased bulk associated with larger connectors likely enhanced resistance to crack initiation and propagation within areas subjected to elevated tensile stress, particularly at the distal abutment wall and cantilever connector region. Consequently, connector design should be considered a critical factor during treatment planning and restoration design for implant-supported cFPDs.

The fracture patterns observed in the present study further support the mechanical findings. All specimens demonstrated catastrophic fracture of the zirconia framework without deformation or fracture of the implant analogs or abutment screws, indicating that failure was confined primarily to the ceramic structure. Crack propagation consistently originated at the distal abutment wall and extended toward the connector region, suggesting that this region represents a critical zone of stress concentration during functional loading. Furthermore, SEM revealed the presence of compression curls within the connector region, supporting the role of localized tensile stress in crack initiation and propagation. Similar fracture patterns have been reported in previous investigations evaluating zirconia cantilever restorations, further confirming the susceptibility of the connector region to tensile stress accumulation [12].

To simulate clinically relevant implant positioning, a 2° mesiodistal angulation was applied to the premolar implant analog, while the molar analog remained parallel to the long axis. This configuration reflects common clinical limitations encountered in posterior mandibular implant placement and improved analog stability during fabrication and mechanical testing. Furthermore, the use of engaging and non-engaging titanium bases was based on previous evidence demonstrating improved rotational stability and stress distribution in cantilevered implant-supported restorations [29]. Thermocycling was performed to simulate approximately half year of clinical service and reproduce thermal fluctuations encountered within the oral environment [30]. These methodological considerations may enhance the clinical relevance of the present findings.

The findings of the present study have important clinical implications regarding the selection of zirconia materials and connector design for implant-supported cFPDs. Monolithic 3Y-TZP zirconia with larger connector dimensions may be more suitable for posterior restorations and patients with increased occlusal loading because of its superior fracture resistance. Conversely, multilayer zirconia materials may be advantageous in esthetically demanding situations where enhanced translucency is desired, provided that connector dimensions are carefully optimized to compensate for reduced mechanical properties.

Several limitations should be considered when interpreting the findings of this study. Static load-to-failure testing does not fully replicate the complex cyclic and multidirectional loading conditions present intraorally. In addition, only 2 connector dimensions and selected zirconia materials were evaluated. Future investigations should assess long-term fatigue behavior under dynamic loading conditions, evaluate alternative connector geometries, and investigate additional zirconia generations and

restorative materials. More extensive thermomechanical aging protocols may also provide a more comprehensive understanding of the long-term clinical performance and durability of implant-supported zirconia cFPDs.

CONCLUSIONS

Within the limitations of this study, both material type and connector dimensions significantly affected the fracture strength of implant-supported cFPDs. Larger connectors and monolithic zirconia still showed superior performance, withstanding forces above typical posterior occlusal loads, underscoring the importance of selecting optimal material and design for long-term clinical success.

Declarations

Declarations of interest: None.

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Author Contribution

N.A. - Conceptualization, Methodology, Investigation, Data collection and writing original draft
H.K. - Validation, Investigation, and Supervision
A.O. - Methodology, and Data collection
S.M.- Methodology
R.L. - Conceptualization, Validation, Supervision, Data analysis, Project administration and Editing

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Data Availability

All data supporting the findings of this study are available within the manuscript.

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Figures

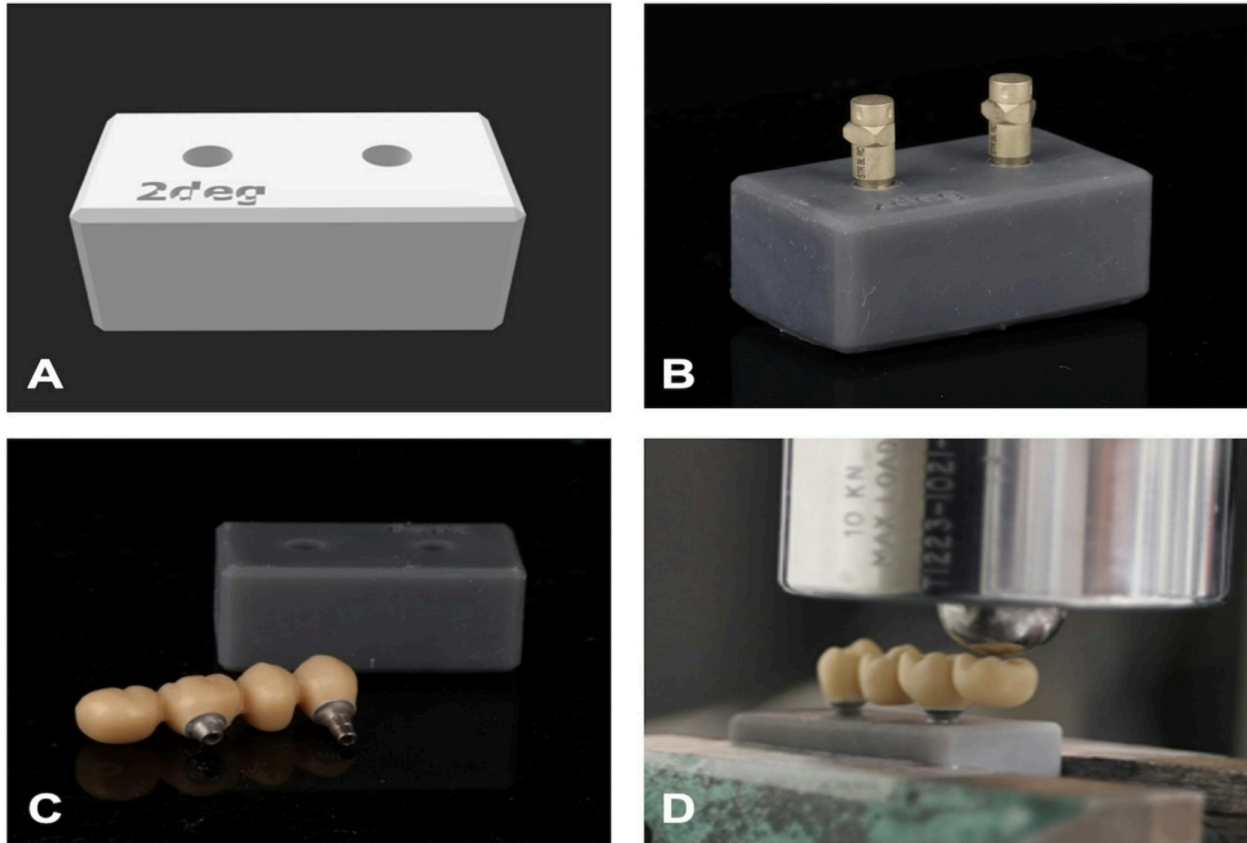


Figure 1

Samples fabrication and testing. A, Digitally designed resin model illustrating implant analog placement. B, Elos scan bodies attached to implant analogs. C, Implant-supported cFPD cemented to titanium bases. D, The assembled specimen was securely fastened to the universal testing machine for mechanical load application and fracture strength assessment.

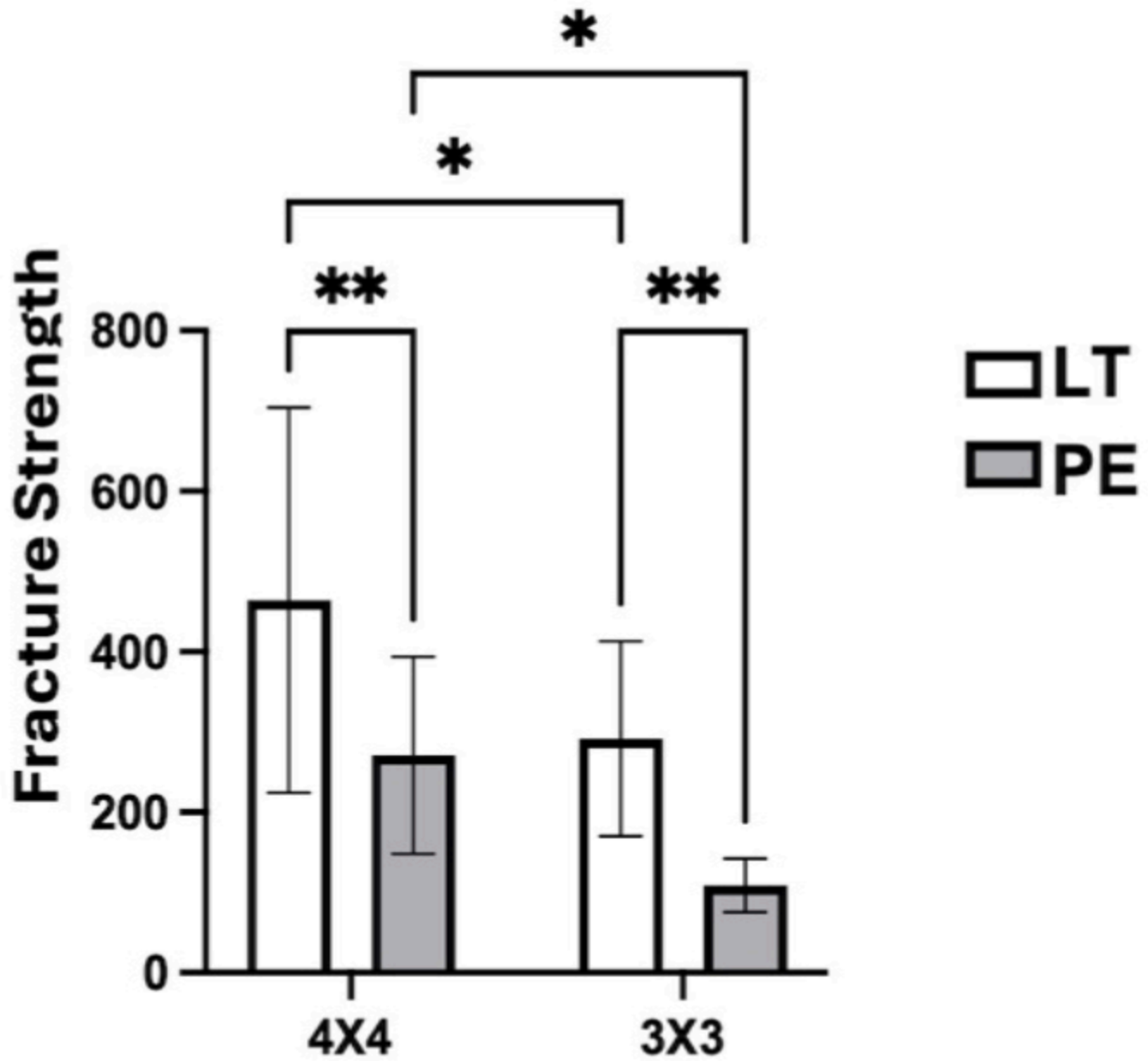


Figure 2

Bar graph of mean values of fracture strength (N).

* $p < .05$; ** $p < .01$.

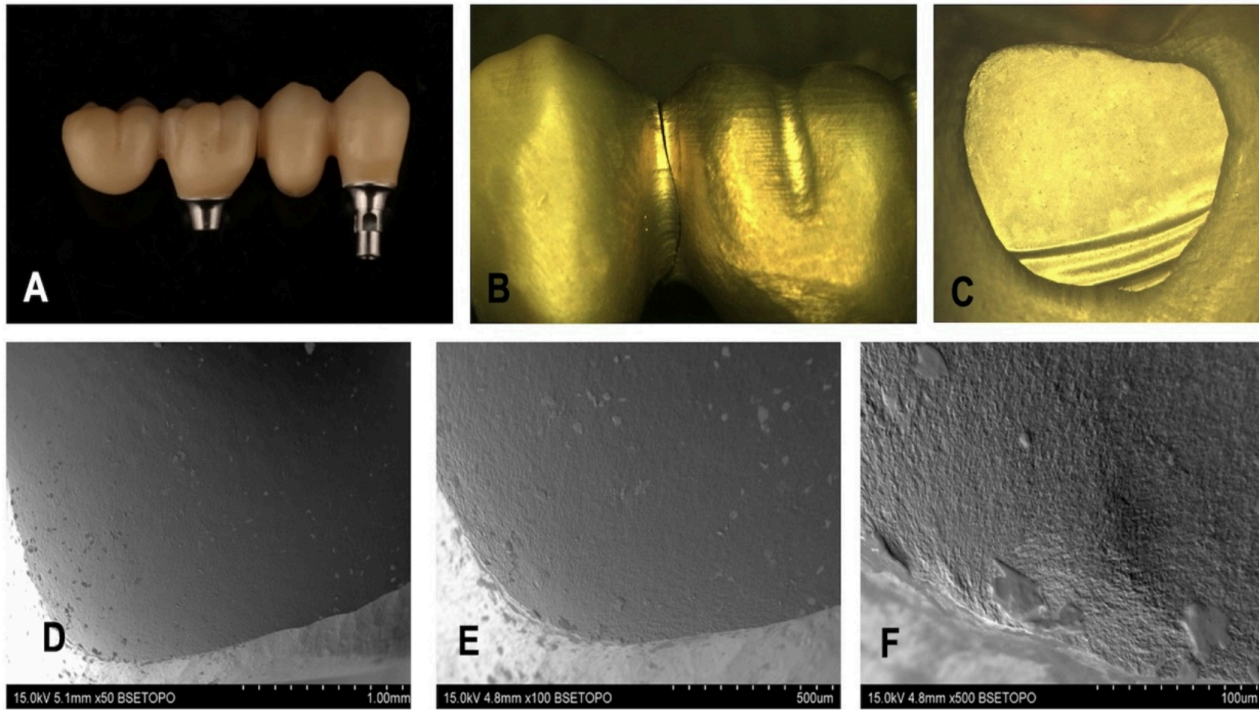


Figure 3

Fracture Pattern. A, Typical fracture of Implant-supported cFPD propagating from the distal abutment wall. B, Representative Olympus SZX12 stereo microscope image demonstrating fracture in the distal abutment propagating through the connector region. C, Representative Olympus SZX12 stereo microscope image demonstrating fracture through the connector region. D-F, SEM images illustrating fracture propagation in the distal abutment wall through the coronal aspect of the connector region. The side indicating the direction of load application is marked. D, ×50 magnification. E, ×100 magnification. F, ×500 magnification.