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Fracture resistance of fixed partial dentures: the influence of restoration geometry and material in additive manufacturing

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PURPOSE. The location of the edentulous area in the dental arch can influence the design of the bridge prosthesis in the surrounding region and the forces it will encounter. This study assessed the fracture strength of restorations with various geometric designs produced using different additive and subtractive manufacturing methods. **MATERIALS AND METHODS.** Co-Cr metal and zirconia fixed partial denture (FPD) frameworks were designed in both linear and curved geometries. The Co-Cr metal frameworks were produced through casting (C) and laser sintering (L), while the zirconia (Z) frameworks were obtained through milling (n = 10). After veneering the frameworks, a four-point bending test was conducted on the specimens to assess their fracture strength. All obtained values were statistically analyzed ($P < .05$). **RESULTS.** In both linear and curved groups, Z group showed the lowest fracture resistance values followed by C and L groups and the differences between the groups were found statistically significant ($P < .05$). In L group, curved FPDs showed statistically significantly higher fracture resistance values than linear FPDs ($P < .05$). In both Z and C groups, curved FPDs showed statistically significantly lower fracture resistance values than linear FPDs ($P < .05$). **CONCLUSION.** The geometric configuration of the restoration and manufacturing technique affects the fracture resistance of different framework materials in FPDs. [J Adv Prosthodont 2025;17:92-100]

KEYWORDS
Additive manufacturing; Fracture resistance; Framework design; Zirconia

INTRODUCTION

Dental porcelains provide significant benefits as dental materials due to their biocompatibility, color stability, light transmittance, and resistance to abra-

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sion, with a thermal expansion coefficient that is similar to that of natural teeth.[1] Nevertheless, inherent features such as "Griffith cracks" within their internal structure and shrinkage during firing contribute to their characteristic brittle nature.[2] To enhance the durability of FPD restorations against intraoral forces, strategies involving the incorporation of a metal alloy or alternative ceramic material with superior fracture strength values can be employed to fabricate a framework.[3]

In producing multi-unit FPDs, various framework materials have been used to support the layered veneer ceramics. Today, the most commonly used framework materials are non-precious metal alloys (e.g. cobalt-chromium) or zirconia. Non-precious metal alloys have been produced using the lost wax technique for many years.[4,5] However, the metal casting technique has disadvantages, such as requiring a lot of time in the wax modelling and design phase, having many error-prone laboratory stages, requiring technical precision, and producing less compatible restorations due to shrinkage during casting.[6,7] For such reasons, computer-aided production methods are frequently used in the production of non-precious metal alloys just like zirconia.[4,8]

Computer-aided design (CAD) and computer-aided manufacturing (CAM) systems have been used in many areas of dentistry. All CAD-CAM systems have three main stages: digitalization, design and manufacturing under computer control. CAM systems are divided into two subgroups: subtractive and additive manufacturing methods.[9-11] In the subtractive manufacturing (SM) method, the restoration designed in the software is produced in three dimensions as a result of grinding a whole block with burs that can move in various axes.[12,13] Subtractive production methods have brought many advantages. Patients' waiting times have been shortened, and patient and physician comfort has increased as better restorations can be obtained in a shorter time.[14] Since the blocks to be abraded in this method are prepared in a homogeneous structure in the factory, the physical and microstructural properties of the obtained restorations do not change depending on the technician, and restorations with superior properties can be produced.[9,13]

In the additive manufacturing (AM) method, the designed restoration is manufactured step by step as addition of each layer on the previous one.[15] Additive manufacturing methods have been defined by the American Society for Testing and Materials (ASTM) as "it is a method of combining materials layer by layer to produce objects designed from three-dimensional models.". The working principle of additive methods is based on the manufacturing of a three-dimensional object by printing each section on the previous one after obtaining a series of cross-sectional sections of the designed geometry.[16,17] The primary advantages of AM methods are that objects with complex geometry can be produced without leaving any residual material, high printing accuracy and relatively short production time.[18-20]

One of the important factors affecting the long-term success of FPD and especially all-ceramic systems, is fracture resistance.[21-24] Several studies in the literature compare different framework materials supporting the layered veneer ceramics that represent multiunit restorations.[23,25,26] However, those studies all focused on one restoration design.

The exact location of edentulous areas in the dental arch plays a crucial role in the ideal design of FDPs, thus affecting the mechanical stresses involved. Notably, the geometric structure of FDPs differs based on where they are placed in the mouth; posterior FDPs usually have a straight shape while anterior FDPs are characterized by a curved design. However, a literature review reveals that no studies have examined the fracture strengths of restorations with varying geometries during their production. This study aims to evaluate the fracture strength of restorations with different geometric designs produced using different additive and subtractive manufacturing methods. The null hypothesis is that restoration geometry and manufacturing methods have no effect on the fracture resistance of FPDs.

MATERIALS AND METHODS

The current study involved the preparation of two distinct models mimicking dental prostheses: a four-unit fixed partial denture (FPD) in a posterior straight configuration and a six-unit FPD in an anterior curved

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configuration, utilizing typodont resin models (AG-3; Frasaco GmbH, Tettnang, Germany). In the former model, the maxillary right first premolar and maxillary right second molar teeth were prepared, while in the latter model, the maxillary right canine and the maxillary left canine teeth were prepared, featuring a circumferential chamfer finishing line measuring 1.2 mm, an incisal/occlusal height reduction of 1.5 mm, and a 6° axial inclination. Subsequently, all sharp contours were rounded. The inter-abutment typodont teeth were excised to simulate partial tooth loss in discrete regions.

For the fabrication of metal master models, the prepared resin models were scanned using a high-resolution industrial scanner (SolutionixRexcan CS2+; MEDIT Corp., Seoul, Korea) and were subsequently milled utilizing a CNC device (Best Marlow; BWM 6050, Nagoya, Japan) in accordance with standard tessellation language (STL) data.

Prior to specimen production, a power analysis conducted through G*Power version 3.1.9.7 for sample size estimation based on data from prior research[27] indicated a minimum sample size of 10 specimens to achieve a 5% sampling error, 95% power. The metal master models were digitized using a laboratory scanner (CS.Neo2; CADStar, Bischofshofen, Austria), and the data were then transferred to design software to create a four-unit linear FPD framework and a six-unit curved FPD framework. Coping thickness was set at 0.5 mm for metal-supported restorations and 0.8 mm for zirconia restorations, as per manufacturer recommendations. Marginal cement space was established at 20 µm, with an internal space of 50 µm provided from the upper margin of 0.5 mm. The designed frameworks underwent fabrication utilizing three distinct manufacturing techniques: casting for Co-Cr frameworks, direct metal laser sintering (DMLS) for Co-Cr frameworks and milling for zirconia frameworks. This resulted in a total of 60 frameworks prepared (n = 10 per group, a total of 6 groups).

In the casting groups (C), STL files were transferred to a 5-axis milling machine (HyperDent; Follow Me Technology, Munich, Germany) to obtain wax patterns, which were then cast with Co-Cr alloy (Wirobond 280; BEGO, Bremen, Germany) using a centrifugal apparatus (Fornax T; BEGO, Bremen, Germany). The DMLS groups (L) utilized the same STL files, directing them to a laser sintering device (ProX 100DP; 3D Systems, SC, USA) for fabrication of Co-Cr alloy frameworks. In the zirconia coping groups (Z), frameworks were milled from zirconia blocks (Zirkonzahn ICE Translucent; Zirkonzahn, Brunico, Italy) and subsequently sintered at 1500°C using a sintering furnace (Tegra Speed; Yenadent, Istanbul, Turkey). Veneering was performed with dentin and enamel porcelain (VMK Master dentin and enamel/VM9 base dentin; VITA Zahnfabrik, Bad Säckingen, Germany), followed by glazing (Akzent Plus Glaze; VITA Zahnfabrik, Bad Säckingen, Germany) according to manufacturer

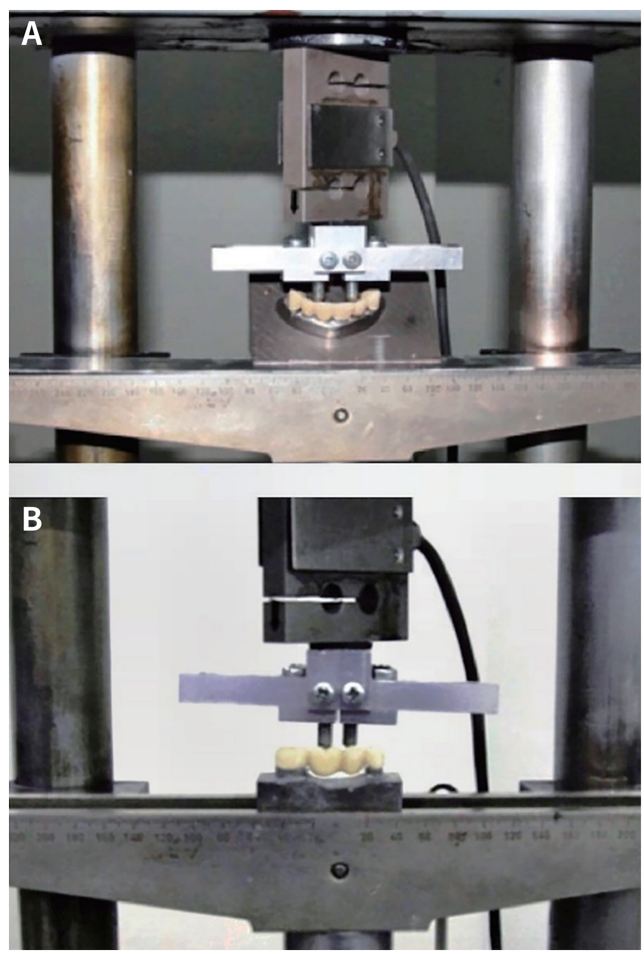

Fig. 1. Four-point bending test was conducted utilizing a universal test device to assess the fracture strength of the specimens. (A) Curved specimens at four-point bending test. The fracture load was applied with 45 N vector on the singular area. (B) Curved specimens at four-point bending test. The fracture load was applied on the occlusal surface.

guidelines. An experienced laboratory technician manufactured all specimens.

Subsequent to veneering procedures, a four-point bending test was conducted utilizing a universal test device (AVK Budapest; Hungary, AVK; Budapest, Hungary) to assess the fracture strength of the specimens (Fig. 1). Prior to testing, specimens were cemented to a metal study model with glass ionomer cement (Ketac; 3M ESPE, St. Paul, MN, USA) following manufacturer recommendations. Steel spheres with a diameter of 4 mm were positioned on the specimens corresponding to predetermined areas during veneering. The universal testing machine was carefully calibrated for precise alignment. The application jaw of the apparatus was meticulously adjusted to ensure that it contacted all steel spheres simultaneously. This adjustment facilitated an even distribution of force. The force was then applied in a controlled, gradual manner, increasing consistently at a rate of 0.5 millimeters per minute. This methodical application enabled accurate measurements of the materials' strength. After each test, the resulting fracture strength values were meticulously documented. These values, reported in Newtons, indicated the force needed to fracture the zirconia-based restorations. For metal-based restorations, the maximum load was measured. This maximum load, also in Newtons, represented the force each specimen could withstand before the supporting framework fractured. The application jaw of the universal testing machine was precisely aligned to contact the steel spheres at the same time, applying force at the rate of 0.5 mm per minute. After testing, Newton values for the fracture strength of zirconia restorations were measured. For maximum load for metal restorations—representing the force the specimens could endure without framework fracture—were documented. All mechanical tests were conducted by a single operator (E.K.C.). Statistical analysis was conducted using R version 2.15.3 software (R Core Team; 2013, Vienna, Austria), employing median, first-quartile, and third-quartile assessments. Normal distribution conformity of quantitative data was assessed using the Shapiro-Wilk test and graphical methods. The Mann-Whitney U test was utilized for evaluating variables displaying non-normal distribution between two groups, while the Kruskal-Wallis test was applied for intergroup evaluations of non-normally distributed variables, supplemented by the Dunn-Bonferroni test to ascertain significance sources. Pearson correlation coefficient was employed to determine quantitative variable relationships, with statistical significance set at $P < .05$.

RESULTS

In straight restorations, there was a statistically significant difference in fracture strength values according to the production method ($P < .001$) (Table 1). As a result of the pairwise comparisons performed using the Dunn-Bonferroni test, the values for both the straight

Table 1. Four-point bending test results of casting metal, laser sintering metal and milled zirconia fixed partial dentures. Results are presented as median (first quartile, third quartile) and in Newton units

Production method	Straight Median (Q1, Q3)	Curved Median (Q1, Q3)	^{b}P
Casting metal (C)	2058 (1985, 2121) N	1911.5 (1845, 1948) N	0.010*
Laser sintering metal (L)	2268 (2010, 2384) N	2633 (2562, 2669) N	< 0.001**
Zirconia (Z)	1005.5 (982, 1084) N	954 (934, 973) N	0.021*
^{c} p (C/L/Z)	< 0.001**	< 0.001**	
p (C/L)	0.999	0.033*	
p (C/Z)	0.001**	0.033*	
p (L/Z)	< 0.001**	< 0.001**	

^{b} Mann-Whitney U test, ^{c} Kruskal-Wallis test.
*$P < .05$, **$P < .01$.

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and curved design Z groups were statistically lower than those of the other groups (P < .001). As a result of the pairwise comparisons performed using the Kruskal-Wallis test, no statistically significant differences were observed in straight designs between group C and group L (P = .999)(Table 1).

In curved restorations, there was a statistically significant difference in fracture resistance values according to the production method (P < .001). As a result of the pairwise comparisons performed using the Dunn-Bonferroni test, the values of the curved design of group Z were significantly lower than those of the curved design of the group C (respectively, P = .033, P < .001). The values of the curved design of group C were found to be significantly lower than the values of the straight design of group C (P = .033). Curved design of the group L was found to be statistically significantly higher than the fracture strength values of the straight design of the group K (P < .001). The straight design of the group Z showed statistically significantly higher fracture strength values than the curved design of group Z (P = .021). A significant reduction in fracture resistance associated with curved geometry was observed for casting metal (P = .010) and zirconia (P = .021) compared to their straight counterparts. Group L demonstrated superior fracture resistance compared to all groups (P < .001) for both geometries (Fig. 2).

DISCUSSION

This study aims to determine the fracture strength of straight and curved FPDs produced with different fabrication techniques and framework materials. For this purpose, two different models were created to represent two cases, and the fracture strengths of straight and curved restorations produced by three methods were examined. The null hypothesis was rejected as the restoration geometry and manufacturing methods affected the fracture resistance of FPDs.

In dentistry, specimens for fracture strength studies can usually be prepared in disk, rod or anatomical tooth forms.[27-29] The present study decided to prepare the specimens in an anatomical form since it was predicted that the fracture values would be closer to *in-vivo* conditions[30] when the specimens were prepared in an anatomical form. When the studies are examined, it is seen that the specimens were ce-

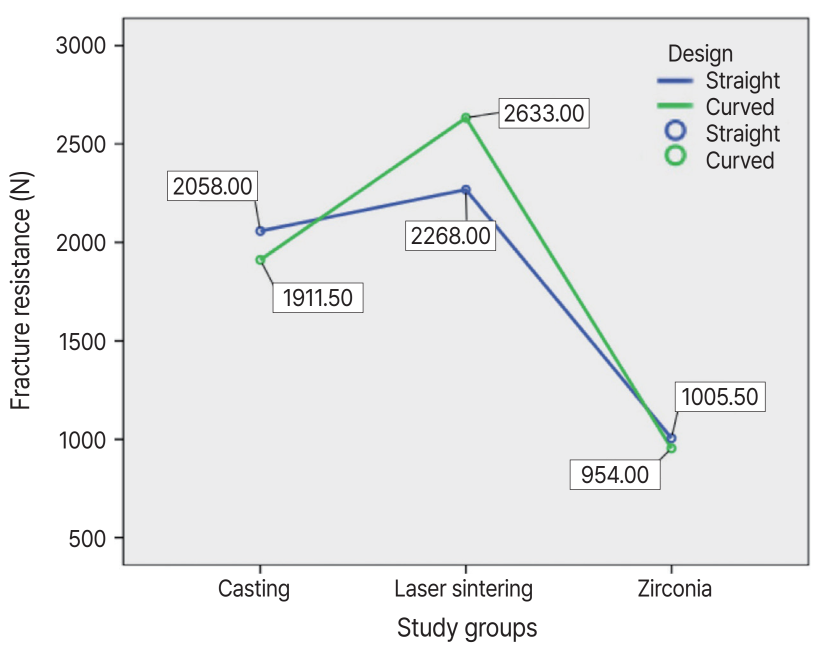

Fig. 2. The line chart of the fracture resistance results of the groups (P < .01).

mented before the tests, with different types of cement such as temporary cement,[31] glass ionomer,[32] resin-modified glass ionomer,[33] resin[34] or zinc phosphate.[35] In this study, glass ionomer cement, which is frequently used in clinical applications of metal and zirconia-based restorations, was preferred and finger pressure was applied during the cementation process to mimic clinical conditions.

In the present study, the four-point bending test was preferred to evaluate the fracture strength of the specimens. It has been reported that the loading speed adjusted while performing the bending tests on universal tests also affects the results. At increasing loading rates, incorrect data can be recorded due to the lack of time for crack growth. For this reason, it is stated that the preferred loading speed should be as low as possible.[36,37] In this study, the loading speed of the samples was determined as 0.5 mm/min.

This study followed a digital workflow to ensure standardization and eliminate personal errors in the process from the measurement stage to the production of frameworks. For the casting group, to minimize the errors caused by the technician, the wax models were cast with the conventional method after the computer-aided milling system was obtained. A skilled technician utilized silicon keys to veneer all specimens and ensure that ceramic processing into the frameworks was standardized.

Studies have shown that the fracture strength values specified for bridge restorations with zirconia framework vary between 425 N and 2009 N. These variations can be caused by many parameters such as the diameter of the joint areas, the length of the restorations, the different universal test methods used, aging or not, cementation preference, the thickness of the framework used, veneering, and the material of the main model.[23,25,26] Since no studies are comparing the fracture strength values of straight and curved bridge restorations among the published studies; findings were evaluated within the study itself (Table 1).

The zirconia framework group showed statistically significantly lower fracture resistance values compared to the metal framework groups produced by casting and laser sintering methods for both straight and curved bridge designs. While there were no significant differences in fracture resistance between the casting and laser sintering groups for straight bridge designs, the laser sintering group demonstrated significantly higher fracture resistance for curved bridge restorations. Conversely, the cast group exhibited statistically higher fracture resistance for straight bridge restorations than curved ones. Furthermore, although no fractures were observed in the metal frameworks of the straight designs, half of the curved specimens exhibited fractures in their metal substructures. This is believed to be attributable to the smaller connection areas in anterior restorations or potential errors made during the conventional production processes.

Laser-sintered curved bridges exhibited superior fracture strength compared to straight bridges, with no observed cohesive fractures in the metal framework. This indicates that computer-aided fabrication offers greater reliability for metal restorations than traditional methods. In addition to this, laser sintering optimizes load distribution in complex designs, making it ideal for stress-bearing prostheses. The precision of laser sintering also minimizes waste, enabling sustainable creation of patient-specific, load-bearing metal parts.

The fracture strength values of zirconia restorations with a straight bridge design were significantly higher than those with curved designs. It is believed that the reason for this is that the framework thickness in the attachment areas is smaller in anterior restorations as compared to posterior ones. The areas where the framework connects to the anchors are crucial for transferring loads in bridge restorations. Enlarging these connection areas can decrease stress concentration when under load.

Turk *et al*.[27] compared the fracture resistance of metal and zirconia-based crown restorations using two different veneering techniques. When examining the results of the metal and zirconia framework groups veneered by the layering method, which is similar to the present study, it was noted that while there was no statistically significant difference between the groups, the fracture resistance values of the metal substructures were higher than those of the zirconia substructures. In contrast, the current study identifies a statistically significant difference between these groups. Although the results demonstrate a certain level of consistency with the previous study, the statistical evidence does not substantiate a mutual support between the findings. In a study conducted by Kılıçarslan *et al*.,[31] the fracture resistance of metal-based three-unit bridge restorations was compared to restorations produced from metal, lithium disilicate, and zirconia-based inlay bridges. The study found no significant difference between the fracture resistance of zirconia and metal framework bridge restorations. However, the average fracture strength values of three-unit metal-framework bridge restorations in the present study were recorded as 2091 N, which is significantly higher than the 1318 N recorded in the previous study. The difference in results may be due to the use of the Ni-Cr metal framework and the three-point bending test in the previous study. In a study on framework fractures, the authors used the manufacturer's guidelines for the zirconia framework, specifying a connector size of at least 9 mm^2 at the abutment-pontic site and 12 mm^2 at the pontic-pontic site.[38] All fractures were observed at the 9 mm^2 connector area, which may have been influenced by the long-span (4-unit) design in the posterior region. Some researchers suggest that caution should be exercised when following the 9 mm^2 connector size.[39,40] The study identified fractures at the premolar-premolar connection in the posterior framework, while unilateral failure often occurred in the anterior framework beginning at the pontic. Both connectors measured 9 mm^2. For long posterior or anterior curved FDPs, maximizing the height of the central connector is recommended when maintaining the heights of the mesial and distal connectors is not possible. When evaluating connector design, it is essential to consider the impact of material selection. Replacing traditional ceramics with more durable alternatives can significantly improve the overall durability of the framework and decrease the likelihood of catastrophic failure. Additionally, paying close attention to occlusal loading patterns is crucial for reducing stress concentrations, especially in extended FPD.

In clinical practice, integrating digital workflows to design and fabricate FDPs can yield enhanced precision in connector size and shape, ultimately leading to improved biomechanical performance. As the field advances, ongoing research into connector morphology and material innovations will continue to inform best practices, ensuring that both function and esthetics meet the demands of contemporary restorative dentistry. The current study has several limitations. First, it was conducted under *in vitro* conditions, and many parameters were set to ideal levels. Additionally, while the horizontal dimensions of the specimens were similar, the number of pontics differed between the straight and curved subgroups. Lastly, the study design did not incorporate any aging procedures, such as mechanical or thermal aging. Given these limitations, future research could benefit from including a more significant sample size and accounting for potential variability in the manufacturing process.

CONCLUSION

Within the study's limitations, results showed zirconia frameworks had the lowest fracture resistance in both designs. In straight bridges, metal frameworks performed similarly, but in curved designs, cast frameworks showed lower fracture resistance. Curvature negatively impacted fracture strength in both metal and zirconia frameworks, except for laser-sintered metal, which benefited from curvature. Laser-sintered metal frameworks exhibited higher fracture resistance than cast frameworks, particularly in curved designs, demonstrating additive manufacturing's clinical advantage. Material and manufacturing process significantly influence FDP mechanical performance, irrespective of geometry.

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