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Effect of connector cross-sectional area on fracture strength of maxillary anterior provisional fixed partial dentures fabricated using CAD/CAM

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Abstract

Background: Provisional fixed partial dentures (PFPDs) are widely used during fixed prosthodontic treatment; however, the effect of fabrication method and connector cross-sectional area on their fracture resistance remain unclear.

Methods: Three fabrication methods were investigated: additive manufacturing (three-dimensional [3D]), subtractive manufacturing (milled [Mi]), and the conventional method (CM). Connector cross-sectional areas of 9.0, 12.0, and 15.0 mm² were designed, and five specimens were fabricated for each condition (n = 5). Each PFPDs was temporarily cemented onto a test model and subjected to fracture testing using a universal testing machine. The fracture mode was visually examined. The fracture load data obtained from the universal testing machine were analyzed using two-way analysis of variance (ANOVA) followed by Bonferroni post hoc tests ($\alpha = 0.05$).

Results: The mean fracture loads (N) at connector cross-sectional areas of 9.0, 12.0, and 15.0 mm² were 295 ± 33, 425 ± 37, and 590 ± 42 for the 3D group; 218 ± 24, 265 ± 41, and 400 ± 52 for the Mi group; and 177 ± 37, 211 ± 31, and 255 ± 18 for the CM group, respectively. Two-way ANOVA revealed significant main effects of both fabrication method and connector cross-sectional area, along with a significant interaction between the two factors ($p < 0.05$).

Conclusions: In maxillary anterior three-unit PFPDs fabricated using each manufacturing method, the fracture load increased with increasing connector cross-sectional area.

Keywords: Provisional fixed partial denture, CAD/CAM, Connector design, Fracture

strength.

1. Background

In fixed prosthodontic treatment involving crowns and fixed partial dentures, provisional restorations protect prepared abutment teeth and maintain the occlusal relationship [1-3]. In the anterior region, provisional restorations reproduce crown morphology to ensure esthetics and serve as diagnostic tools for the definitive prosthesis. These restorations enable evaluation of parameters such as anterior guidance (inclination angle, and range of movement) and the configuration and distribution of connectors [2, 4, 5]. Polymethyl methacrylate (PMMA)-based and resin composite-based materials are the primary used for provisional restorations [6]. PMMA remains widely used because it is relatively inexpensive and easy to manipulate [7]. In recent years, bis-acrylic composite resins and urethane dimethacrylate resins have also gained popularity for provisional restorations [8].

Provisional restorations can be fabricated using three general techniques: the direct technique, in which chemically polymerizing resin is manipulated directly in the oral cavity; the direct–indirect technique, in which a conventional impression is taken, and the resin is pressed into the impression; and the indirect technique, in which the restoration is fabricated extraorally. The direct and direct–indirect techniques offer advantages in terms of immediacy and simplicity, although their fit and durability are often inferior [9]. In contrast, the indirect technique offers higher accuracy, esthetics, and strength but requires more time and cost [1]. Clinical decision-making should involve selecting an appropriate fabrication method according to each case and treatment plan. For provisional

fixed partial dentures (PFPDs) that connect multiple units, the indirect technique has been recommended because intraoral fabrication procedures often become complex and may lead to poor marginal adaptation or misfit [10].

Computer-aided design/computer-aided manufacturing (CAD/CAM) enables precise control of connector configuration and cross-sectional area during the design process, allowing accurate reproduction of the morphology of provisional restorations in the definitive prosthesis. CAD/CAM fabrication includes subtractive manufacturing and additive manufacturing [11, 12]. In the milling method, provisional restorations are fabricated from pre-polymerized resin blocks or discs by milling [13]. The materials used for this method demonstrate superior color stability, fracture resistance, and surface hardness compared to conventional chemically polymerized resins [14, 15]. Furthermore, their mechanical properties remain unaffected by fabrication conditions, although the method is limited in reproducing complex geometries due to the nature of subtractive milling [16, 17]. In contract The additive manufacturing method, provisional restorations are constructed by sequentially stacking and polymerizing resin layers [5]. PMMA-based resins for additive manufacturing exhibit higher fracture strength, flexural strength, and wear resistance than conventional chemically polymerized resins, although they present lower color stability and greater water sorption [18, 19]. While additive manufacturing offers excellent reproducibility of complex shapes, its mechanical properties may vary depending on the build angle and post-curing parameters [20, 21].

Although several studies have investigated the influence of fabrication methods on the mechanical properties of provisional restorations, some have reported lower fracture strength in additive-manufactured resins compared to milled resin [20, 22]. Conversely, other studies have demonstrated that PMMA-based resins for additive manufacturing

exhibit higher fracture strength than those used in milling and conventional fabrication methods [23, 24]. Therefore, further research is needed to clarify these discrepancies.

Furthermore, when evaluating the clinical success of these materials, structural design factors—such as the dimensions of multi-unit restorations—must be considered alongside inherent material properties [25]In the anterior region, where esthetics hold particular importance, the morphology of the provisional restoration and the cross-sectional form of its connectors must harmonize with the gingiva. Although studies have examined the connector design and cross-sectional area of frameworks for anterior zirconia fixed partial dentures [26], limited information exists regarding the influence of connector cross-sectional area on the strength of provisional restorations. Therefore, this study aimed to evaluate the effects of fabrication method and connector cross-sectional area on the fracture load of maxillary anterior three-unit PFPDsfabricated by additive manufacturing (three-dimensional [3D]), subtractive manufacturing (milled [Mi]), and the conventional method (CM). The null hypothesis stated that neither the fabrication method nor the connector cross-sectional area would affect the fracture load.

2. Methods

2.1. Materials

This study used three different methods to fabricate the PFPDs: 3D, Mi, and CM. The material selection for this study was based on flexural strength testing of simple rectangular specimens (2 × 2 × 25 mm), in conjunction with the flexural strength values specified by the manufacturers. The materials and equipment used in this study are summarized in Table 1. The 3D group used a PMMA-based 3D printing resin (Dima Print Denture Teeth, Kulzer GmbH, Hanau, Germany), the Mi group used a PMMA-based resin

disc (M-PM Disc, Merz Dental GmbH, Lütjenburg, Germany), and the CM group used a chemically polymerized PMMA resin (Unifast II, GC Corp., Tokyo, Japan).

2.2. Specimen fabrication

2.2.1. Abutment model

A stainless-steel abutment model was used to simulate a case of a missing maxillary left central incisor, with the maxillary right central incisor and left lateral incisor serving as abutment teeth (Figure 1). The abutment teeth morphologies were designed according to anatomical dimensions [26]: the diameters of the central and lateral incisors were 6.0 mm and 5.2 mm, respectively, and their axial crown heights were 6.8 mm and 5.3 mm. The distance between the centers of the abutment teeth measured 16.1 mm. Each abutment preparation simulated an all-ceramic crown with a 5° axial taper and a 1.0-mm deep chamfer margin. The abutment model was used in this experiment because its dimensions conformed to anatomical standards.

2.2.2. PFPD design

The abutment model was scanned using a laboratory scanner (S-WAVE Scanner D2000, 3Shape, Copenhagen, Denmark), and the PFPDs were designed with CAD software (Dental Manager 2020, 3Shape). The abutment restorations were simplified without anatomical contouring, with a uniform axial wall thickness of 1.5 mm and an occlusal thickness of 2.5 mm (Figure 2) [26]. A cement space of 120 μm was set [27]. The connector width in the mesiodistal direction was set at 1.0 mm, and only the pontic form was designed with an anatomical shape. Each connector cross-section was designed as an isosceles triangle with rounded corners, with cross-sectional areas of 9.0, 12.0, and 15.0 mm^2 in geometrically similar proportions (Figure 3) [26, 28].

2.2.3. PFPD fabrication

For the 3D group, the designed PFPDs were converted into stereolithography (STL) format and imported into the CAM software (Cara Print CAM 2.0, Kulzer). The build orientation was set at 45°, and the layer thickness at 50 μm. Support structures were designed to avoid the connector area. The PFPDs were fabricated using the 3D printer (Cara Print 4.0 pro, Kulzer) after stirring the resin for 5 min. Post-processing included washing with isopropanol in a cleaning unit (Cara Print clean, Kulzer) and light polymerization for 15 min using a post-curing unit (HiLite power 3D, Kulzer) with a wavelength of 390–540 nm.

For the Mi group, the STL data were imported into the CAM software (GO2dental version 6.09, Shofu Inc., Kyoto, Japan) and milled using a 5-axis milling machine (DWX-53D, Roland DG, Shizuoka, Japan) equipped with three milling burs (BE-0.6-4-DLC, BE-1.0-4-DLC, and BE-2.0-4-DLC, Shofu Inc.).

For the CM group, silicone impression material (EXAFINE, GC Corp) was used to take an index using the 3D group as a reference PFPD, and the restorations were fabricated using the pour-type technique with the manufacturer's recommended powder-to-liquid ratio.

Five specimens were fabricated for each connector cross-sectional area in each group (n=5). The sample size was determined a priori using statistical power analysis software (GPower; Heinrich Heine University Düsseldorf, Düsseldorf, Germany), assuming a significance level of 0.05, a power of 0.8, and an effect size of 0.8. A post hoc power analysis was subsequently performed using GPower to evaluate the achieved statistical power based on the effect sizes calculated from the experimental results.

2.3. Test mold

The test mold is shown in Figure 4. The apical portions of abutments were rounded,

and a 0.4-mm thick layer of silicone impression material (FUSION II, GC Corp) was interposed between each abutment and the holder to simulate the periodontal ligament. Each abutment was embedded 12.0 mm into the holder.

2.4. Temporary cementation of PFPDs

Each PFPD was temporarily cemented to the test mold using a temporary luting carboxylate cement (Hy-Bond Carbo Cement, Shofu Inc.) and stored in distilled water for 24 h before testing.

2.5. Fracture test

The fracture load was measured using a universal testing machine (AG-1 20 kN, Shimadzu Corp., Kyoto, Japan). A hemispherical steel indenter (2.0 mm in diameter) applied the load at an angle of 135° to the long axis of the abutment teeth [26, 29]. The load was applied 3.0 mm from the incisal edge at the midpoint of the pontic in the mesiodistal direction (Figure 5). The crosshead speed was 1.0 mm/min, and the fracture load (N) was recorded at the point of failure.

2.6. Observation of fracture mode

Observations of fracture patterns were performed as supplementary information to aid clinical interpretation of fracture strength results. Fractured specimens were visually examined, and the fracture sites and number of fragments were recorded. Subsequently, the specimens were ultrasonically cleaned in distilled water for 10 min and sputter-coated with palladium-gold. Fracture surfaces were observed using a scanning electron microscope (SEM; SU6600, HITACHI, Tokyo, Japan) at an accelerating voltage of 1.5 kV and a magnification of ×25.

2.7. Statistical analysis

Statistical analysis was performed using SPSS software (Version 29, IBM Corp.,

185 Armonk, NY, USA). Two-way analysis of variance (ANOVA) was conducted with
186 fabrication method and connector cross-sectional area as factors, followed by Bonferroni
187 multiple comparison tests. The significance level was set at 5%.
188
189 **3. Results**
190 The results of the two-way ANOVA are summarized in Table 2. Significant main effects
191 were identified for both fabrication method and connector cross-sectional area, and the
192 interaction between these two factors was also significant ($p < 0.05$).
193 Figure 6 illustrates the fracture load values of PFPDs for each material. For connector
194 cross-sectional areas of 9.0, 12.0, and 15.0 mm², the mean fracture loads were 177 ± 37
195 N, 211 ± 31 N, and 255 ± 18 N for the CM group; 218 ± 24 N, 265 ± 41 N, and 400 ± 52
196 N for the Mi group; and 295 ± 33 N, 425 ± 37 N, and 590 ± 42 N for the 3D group,
197 respectively. The 3D group demonstrated significant differences among all connector
198 cross-sectional areas. The Mi group showed a significant difference only at the 15.0 mm²
199 condition, whereas the CM group displayed no significant differences among connector
200 sizes.
201 Figure 7 Shows the fracture load values for each connector cross-sectional area across
202 different materials. For connector areas of 9.0 mm² and 12.0 mm², the 3D group exhibited
203 significantly higher fracture loads than the other materials. For 15.0 mm², significant
204 differences were observed among all materials.
205 Representative fractured PFPDs are shown in Figure 8. Table 3 shows that in the Mi
206 and CM group, two-piece fractures were frequently observed in the mesial connector
207 region, whereas in the 3D group, multi-fragment fractures were frequently observed in
208 both the mesial connector region and the pontic region.

SEM observations are presented in Figure 9. In all PFPDs, an area surrounded by radially propagating crack lines was observed on the labial-gingival side. The 3D group exhibited smooth fracture surfaces with delaminated layers and fine, evenly distributed arrest lines on the palatal side. The Mi group showed slightly rough but dense fracture surfaces with coarse and evenly spaced arrest lines on the palatal side. The CM group demonstrated rough fracture surfaces with numerous voids and irregularly distributed arrest lines on the palatal side.

4. Discussion

This study tested the null hypothesis that neither fabrication method nor connector cross-sectional area would affect the fracture load of PFPDs. The results showed no significant differences in fracture load among connector cross-sectional areas in the CM group. In contrast, the connector cross-sectional area significantly influenced fracture load in the other fabrication methods, leading to partial rejection of the null hypothesis.

4.1. Test specimens

PFPDs were fabricated using three methods: 3D, Mi, and CM. Although all materials used in this study were primarily composed of PMMA, differences in polymerization mechanisms, polymerization conditions, and internal material structures among the fabrication methods may have resulted in distinct mechanical properties.

In the CM group, provisional restorations were fabricated by mixing and pouring chemically polymerized resin, a process that depends heavily on operator skill. Variations in the powder-to-liquid ratio and manipulation during mixing and pouring may lead to inconsistencies in connector geometry, and incorporation of air bubbles has been reported to reduce fracture strength [30, 31]. Furthermore, this fabrication method is associated

with a decreased degree of polymerization and increased residual monomer content.

In contrast, the PMMA materials used in the Mi and 3D groups are manufactured under industrially controlled conditions [32], which are expected to provide a higher degree of polymerization and a more homogeneous internal structure. Milled PMMA blocks are polymerized under high temperature and pressure, resulting in increased molecular weight and enhanced intermolecular bonding, which may contribute to improved mechanical properties.

Additively manufactured PMMA-based materials are light-polymerized resins fabricated in a layer-by-layer manner, which inherently results in a laminated structure. This structural characteristic may introduce anisotropy, and the interlayer bonding strength may influence fracture behavior depending on the loading direction [33]. Nevertheless, the ability to reproduce highly accurate and standardized connector designs may represent a distinct advantage over the conventional method. Furthermore, when printing parameters such as build orientation, layer thickness, and post-curing conditions are appropriately optimized, additively manufactured PMMA-based restorations may achieve fracture strength comparable to that of milled PMMA restorations.

CAD/CAM fabrication techniques enable consistent and standardized reproduction of restoration morphology [19, 34]. Therefore, besides improved shape reproducibility, the intrinsic homogeneity of CAD/CAM PMMA materials may have contributed to the stability of fracture strength observed in the Mi and 3D groups.

Stainless-steel abutments were selected to maintain consistent abutment geometry. However, stainless steel possesses a higher elastic modulus than natural teeth, resulting in higher fracture load values compared to intraoral conditions [30]. To simulate the biomechanical behavior of the periodontal ligament and achieve stress distribution similar

to that of the oral environment, a thin layer of silicone impression material was interposed between the abutments and holder, serving as an artificial periodontal ligament. Finite element analyses have demonstrated that such models produce stress distributions comparable to those occurring *in vivo* [35]. Therefore, incorporating the artificial periodontal ligament was considered effective in creating a testing environment that closely approximates clinical conditions.

4.2. Test conditions

From a mechanical perspective, connector design must provide sufficient strength to withstand bite forces. Previous studies have reported that an isosceles triangular connector with the cervical side as the base demonstrates the greatest mechanical strength [26]. However, in the anterior region, esthetics, cleansability, and harmony with the gingiva must also be considered, and the largest connector area does not necessarily offer the best outcome [36]. Excessive connector cross-sectional area can reduce the interdental space and create over-contoured gingival emergence profiles, resulting in poor esthetics and unfavorable periodontal response [37]. For these reasons, connector cross-sectional areas of 9.0, 12.0, and 15.0 mm^2 were selected to evaluate the balance between strength and esthetics. Although a static fracture test was employed, this approach is considered appropriate for simulating sudden overload conditions in clinical scenarios [38]. Oblique loading typically decreases fracture strength [39]. However, in this study, the load was applied at a 135° angle relative to the long axis of the abutment teeth to simulate the clinical occlusal contact between the maxillary and mandibular anterior teeth.

4.3. Test results

This study found that the 3D group and Mi group exhibited higher fracture load values than the CM group. This is attributed to the homogeneous internal structure obtained via

CAD/CAM and the precise, standardized manufacturing process, which suppresses heterogeneity caused by operator manipulation and air bubbles [24, 40]. The 3D group exhibited higher fracture load values than the Mi group, suggesting that the printing angle during additive manufacturing may be a contributing factor. Reports indicate that the fracture load values of 3D-printed materials vary depending on the manufacturing angle, with higher load values observed at a 45° building orientation [18, 41]. In this study, the 3D group specimens were fabricated at 45°, supporting the influence of build angle on fracture strength. Regardless of the manufacturing method, an increase in fracture load values was observed with increasing connector cross-sectional area. Reports indicate that increasing the connector cross-sectional area of the final prosthesis also results in higher fracture load values [42]. These results suggest that the manufacturing method and the connector cross-sectional area significantly influence the fracture strength.

The maximum bite force in the anterior region has been reported to range from 250–300 N [43, 44]. Comparison of fracture load values showed that the 3D group exhibited the highest values, ranging from 295 N to 590 N across all connector cross-sectional areas. The Mi group showed fracture loads ranging from 218 N to 400 N, suggesting that maintaining a connector cross-sectional area of 12.0 mm^2 or larger may provide sufficient resistance to bite force. In contrast, the CM group showed lower values ranging from 177 N to 255 N, implying that a connector cross-sectional area of at least 15.0 mm^2 would be required to withstand occlusal forces.

Fracture patterns revealed multiple fractures in the mesial connector and pontic regions of the 3D group, whereas the Mi and CM groups showed two-piece fractures localized to the mesial connector. This difference may reflect the interlayer structure and stress distribution specific to additive manufacturing [45, 46]. In the 3D group, fracture

propagated along the layer structure from the mesial connector to the pontic, resulting in multiple fracture sites, whereas stress concentration in the Mi and CM groups led to localized fracture. Two-piece fractures allow easier repair and immediate esthetic restoration, whereas multiple fractures require remanufacturing and may temporarily compromise appearance. Therefore, the 3D group, despite higher fracture loads, may present challenges in repair, whereas the Mi and CM groups may offer advantages in repairability. Optimization of build angles to minimize interlayer delamination and multiple fractures represents an additional important consideration [18].

SEM observations revealed that fracture origins were located on the labial-gingival side in all groups, with radial crack propagation. Previous studies have reported that regions surrounded by radially propagating crack lines may represent fracture initiation sites [42]. This observation is consistent with the findings of the present study. The 3D group exhibited interlayer delamination, whereas the Mi group showed dense fracture surfaces, and the CM group displayed multiple voids along with dense surrounding fracture surfaces. Previous studies have similarly reported smooth surfaces and interlayer delamination in the 3D group and rough fracture surfaces in the Mi group [22]. Despite consistent fracture origins and radial crack propagation across all groups, the observed differences in fracture surface morphology suggest that these variations may be attributed to the fabrication method.

4.4. Clinical implications

Based on the results of this study, the 3D group demonstrated sufficient mechanical strength, indicating a viable option for provisional restorations in the anterior region. The Mi group may also be clinically applicable if the connector cross-sectional area is adequate; specifically, maintaining an area of 12.0 mm^2 or greater is considered sufficient

to withstand occlusal loading. In contrast, the CM group could be used clinically with a connector area of at least 15.0 mm²; however, this design presents limitations in terms of esthetics and gingival harmony.

Furthermore, differences in fracture patterns suggest that repairability may be more favorable in the Mi and CM groups [47]. Therefore, clinical selection should consider both fracture strength and ease of repair and remanufacturing. In the 3D group, optimization of printing parameters could improve fracture patterns and expand the clinical applicability of digitally fabricated PFPDs.

4.5. Study limitations

This study employed static fracture testing using stainless-steel abutments, which did not fully replicate natural teeth or intraoral conditions. Long-term factors such as fatigue loading, thermal cycling, and the salivary environment were not considered. Future studies should include natural teeth, dynamic fatigue testing, and long-term clinical simulations. Additionally, the simplification and standardization of abutment morphology and connector cross-sectional area limited the ability to fully reproduce clinical occlusal relationships, representing an inherent clinical limitation of this study.

5. Conclusions

In the three fabrication methods used in this study, fracture load increased with increasing connector cross-sectional area.

Fracture loads sufficient to withstand clinical occlusal forces were achieved at connector cross-sectional areas ≥9.0 mm² for the 3D group, ≥12.0 mm² for the Mi group, and ≥15.0 mm² for the CM group.

List of abbreviations

PMMA — Polymethyl methacrylate

PFPDs — Provisional fixed partial dentures

CAD/CAM — Computer-aided design/computer-aided manufacturing

3D — Three-dimensional

Mi — Milled

CM — Conventional method

STL — Stereolithography

SEM — Scanning electron microscope

ANOVA — Analysis of variance

Declarations

Acknowledgments

Not applicable.

Authors' contributions

MK and TS conceived and designed the study, fabricated the specimens, performed the experiments, analyzed and interpreted the data, and drafted the manuscript. SN performed

the statistical analyses and contributed to data interpretation. HS critically reviewed the manuscript and provided intellectual input and revisions. All authors read and approved the final manuscript.

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Availability of data and materials

The datasets used and/or analyzed during the current study are available from the corresponding author on reasonable request.

Ethics approval and consent to participate

Not applicable.

Consent for publication

Not applicable.

Competing interests

The authors declare that they have no competing interests.

Clinical trial number

Not applicable.

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561

Table 1. Materials and equipment used in this study.

Materials	Product name	Manufacturer	Lot No.
Polymethyl methacrylate (PMMA)–based 3D printing resin	dima Print Denture Teeth	Kulzer GmbH, Hanau, Germany	AD22F13A3
Polymethyl methacrylate (PMMA)–based resin disc	M-PM Disc	MERZ Dental GmbH, Lütjenburg, Germany	20322
Polymethyl methacrylate (PMMA)	Unifast II	GC Corp., Tokyo, Japan	2110191

562

This table lists the polymethyl methacrylate (PMMA) used in the manufacture of provisional fixed partial denture. Product name, manufacturing method, equipment used, and lot number are included.

563

Table 2. Results of two-way analysis of variance for fracture load values.

Source of variation	Type III Sum of Squares	df	Mean Square	F-value	P-value
Fabrication method	408,480.69	2	204,240.35	128.090	< 0.01
Connector cross-sectional area	245,835.31	2	122,917.66	77.088	< 0.01
Interaction	72,308.87	4	18,077.22	11.337	< 0.01

564 Statistically significant differences are indicated by $P < 0.01$.

565

Table 3. Results of fracture location in destructive testing.

CM group			
	9.0	12.0	15.0
Mesial Connecter	5	5	4
DistalConnecter	0	0	1

Both Connecter	0	0	0
Mi group			
	9.0	12.0	15.0
Mesial Connecter	5	3	3
Distal Connecter	0	2	2
Both Connecter	0	0	0
3D group			
	9.0	12.0	15.0
Mesial Connecter&Pontic	3	5	2
Distal Connecter&Pontic	1	0	1
Both Connecter &Pontic	1	0	2

Results of classification of fracture locations in destructive testing according to manufacturing method and connector cross-sectional area.

566

567 **Figure legends**

568 **Figure1.** Dimensions of the abutment mold.

569 Abutment tooth morphologies were designed according to anatomical dimensions.

570

571 **Figure2.** Dimensions of provisional fixed partial denture.

572 The abutment structure for a provisional fixed partial denture was simplified rather than

573 anatomically shaped, whereas the pontic structure was designed with an anatomical form.

574

Figure3. Schematic of the connector cross-section.

The cross-sectional area of the joint was designed to be similar for each of the three conditions: 9.0, 12.0, and 15.0 mm^2.

Figure4. Dimensions of the test mold.

The mold was made of stainless steel, with the apex of the abutment tooth designed in a rounded shape. A simulated periodontal ligament 0.4 mm thick was interposed between the abutment tooth and the holder.

Figure5. Schematic diagram of the destructive test setup.

Left: lateral view of the fracture test setup. Right: loading point on the pontic.

Figure6. Comparison of connector cross-sectional area by manufacturing methods.

Differences in letters indicate significant differences ($p < 0.05$).

Figure7. Comparison of manufacturing methods for connector cross-sectional area.

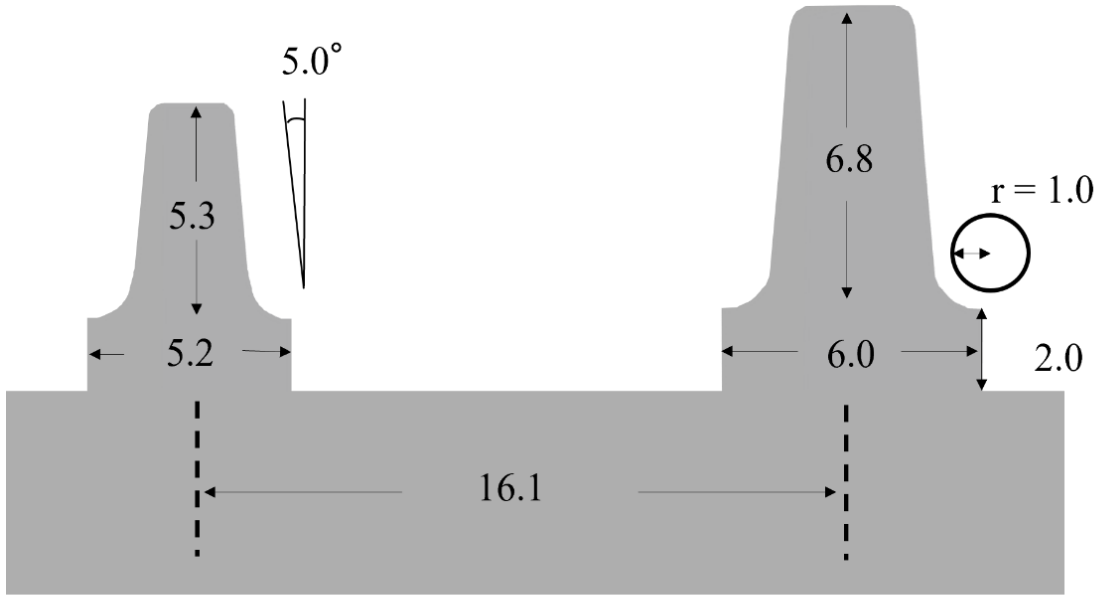
Differences in letters indicate significant differences ($p < 0.05$).

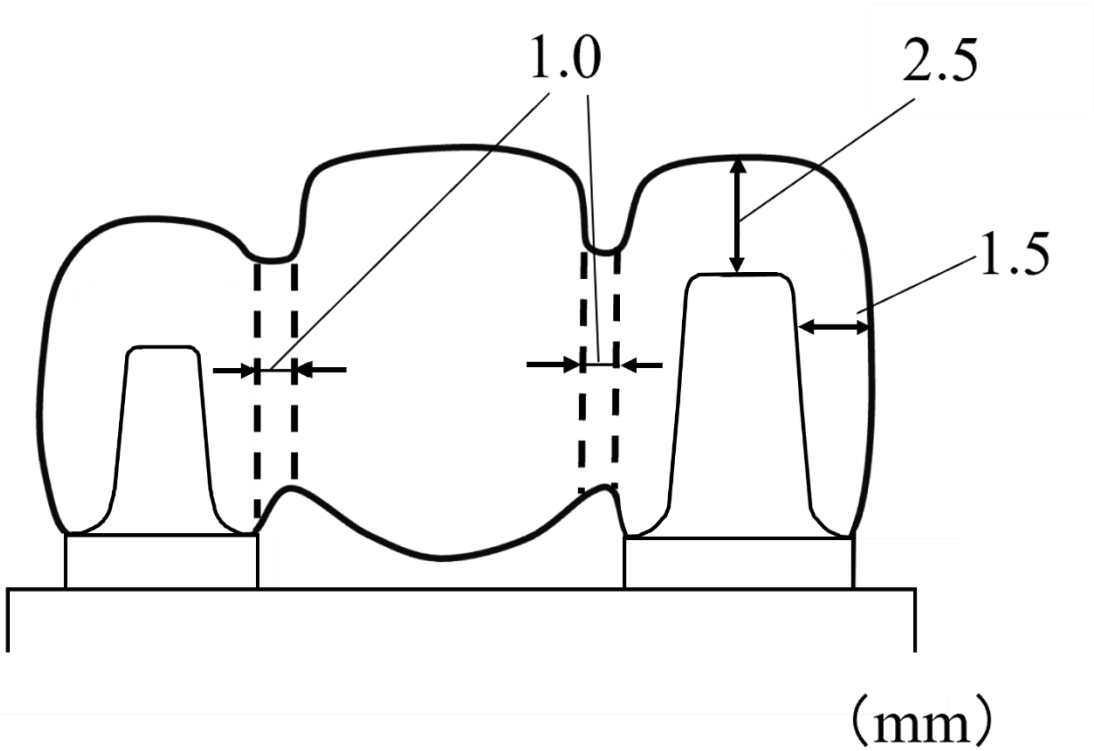
Figure8. Visual inspection of provisional fixed partial denture fracture patterns.

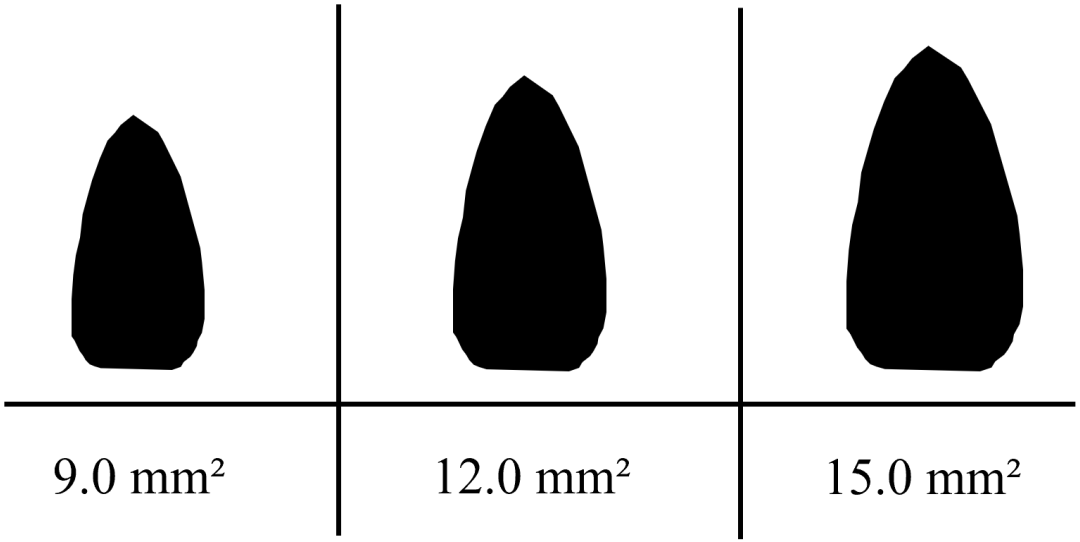
Representative fractured specimens from each fabrication group are shown. All specimens are oriented with the lingual surface facing the frontal direction, corresponding to the loading point, to standardize observation of fracture location and morphology.

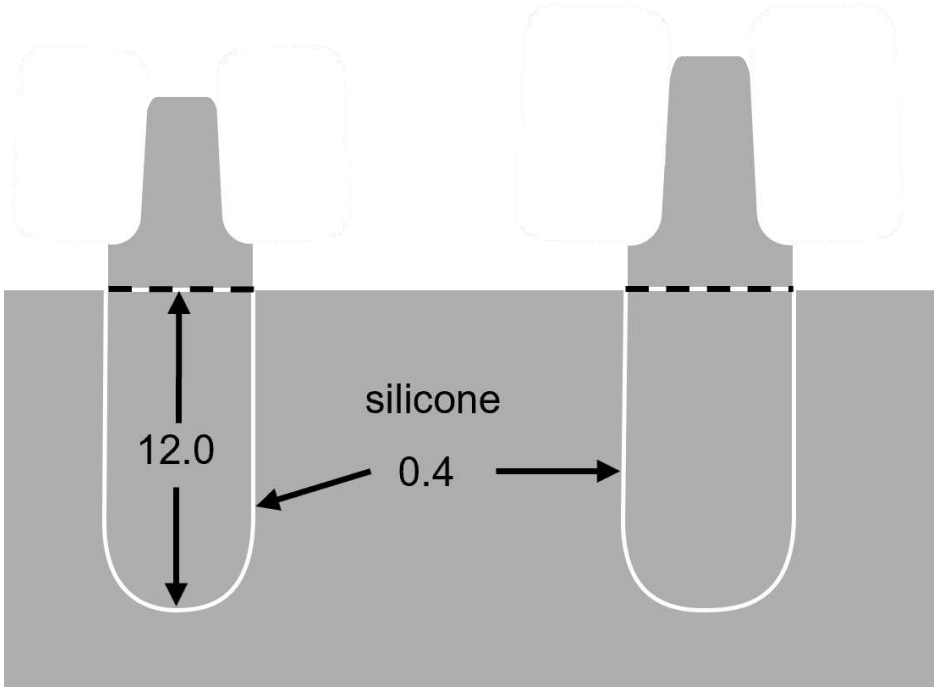
Figure9. Scanning electron microscopy (SEM) images of the fracture surface after destructive testing.

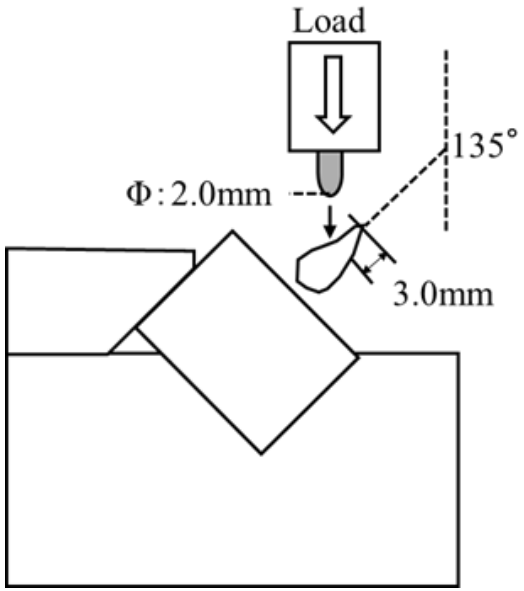
A schematic diagram of the connector area observed by SEM is shown. Arrowheads indicate regions enclosed by radially propagating crack propagation lines, suggesting the fracture origin.

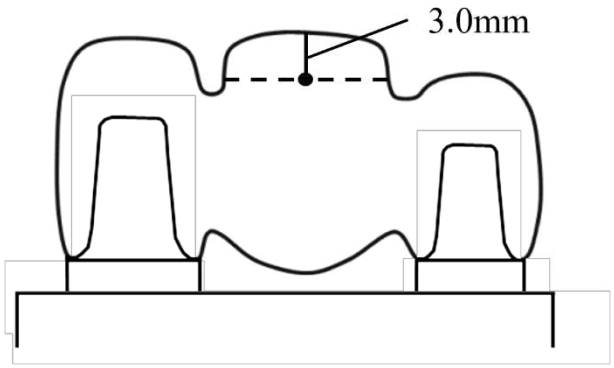

5.0°
5.3
5.2
6.8
r = 1.0
6.0
2.0
16.1
(mm)

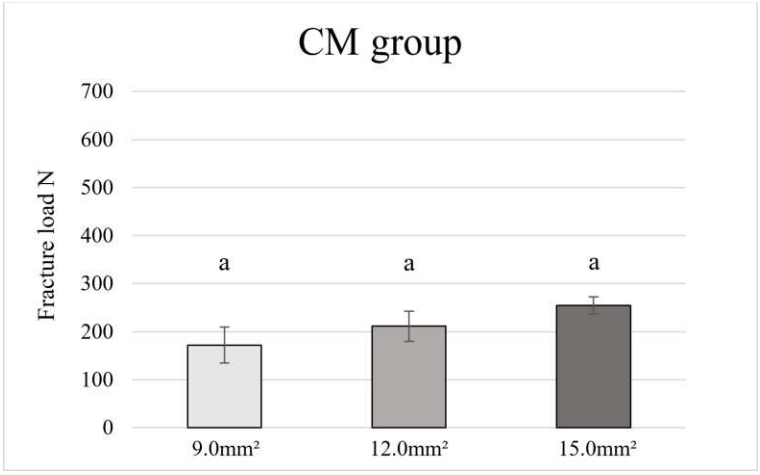

1.0
2.5
1.5
(mm)

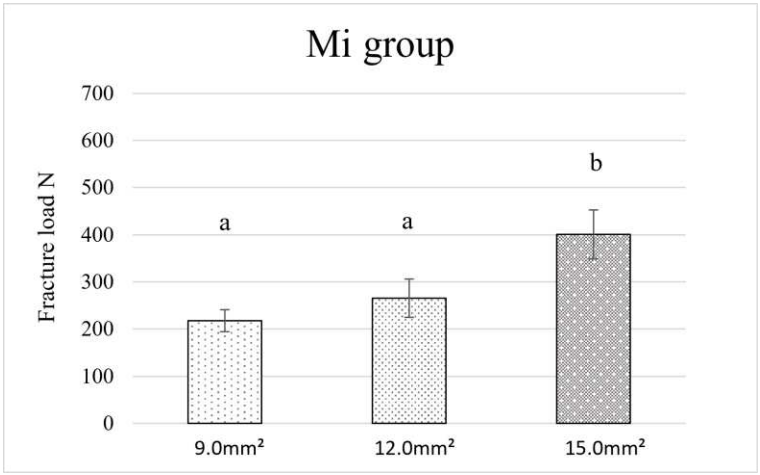

9.0 mm²
12.0 mm²
15.0 mm²

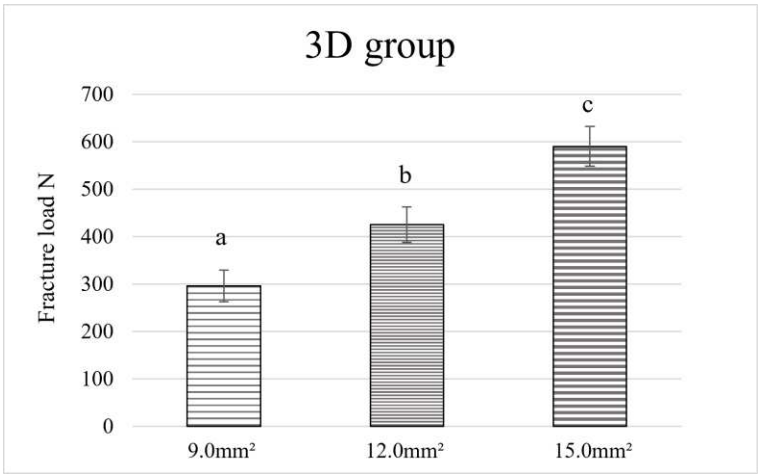

12.0
silicone
0.4

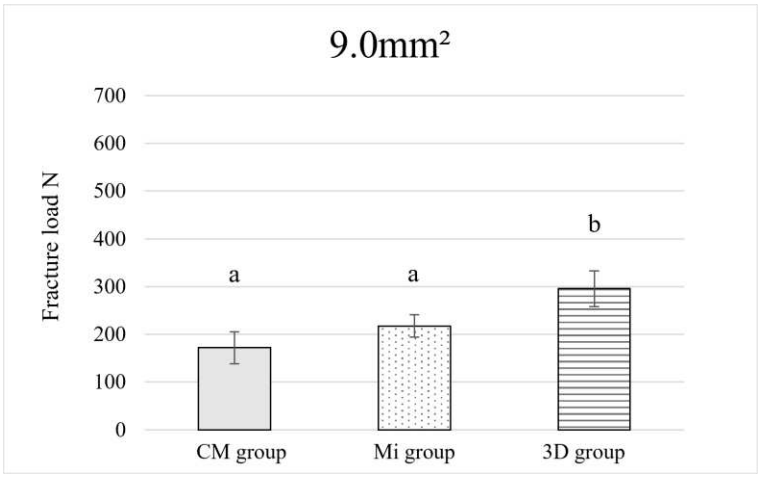

Load
135°
Φ:2.0mm
3.0mm

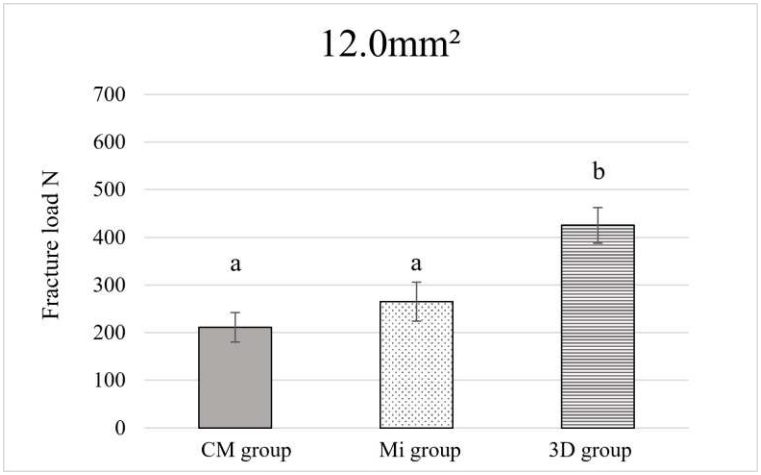

3.0mm

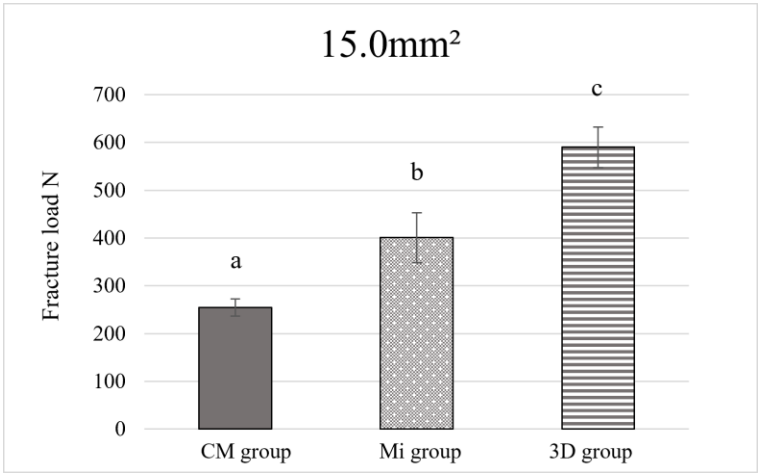

CM group
Fracture load N
700
600
500
400
300
200
100
0
a
a
a
9.0mm²
12.0mm²
15.0mm²

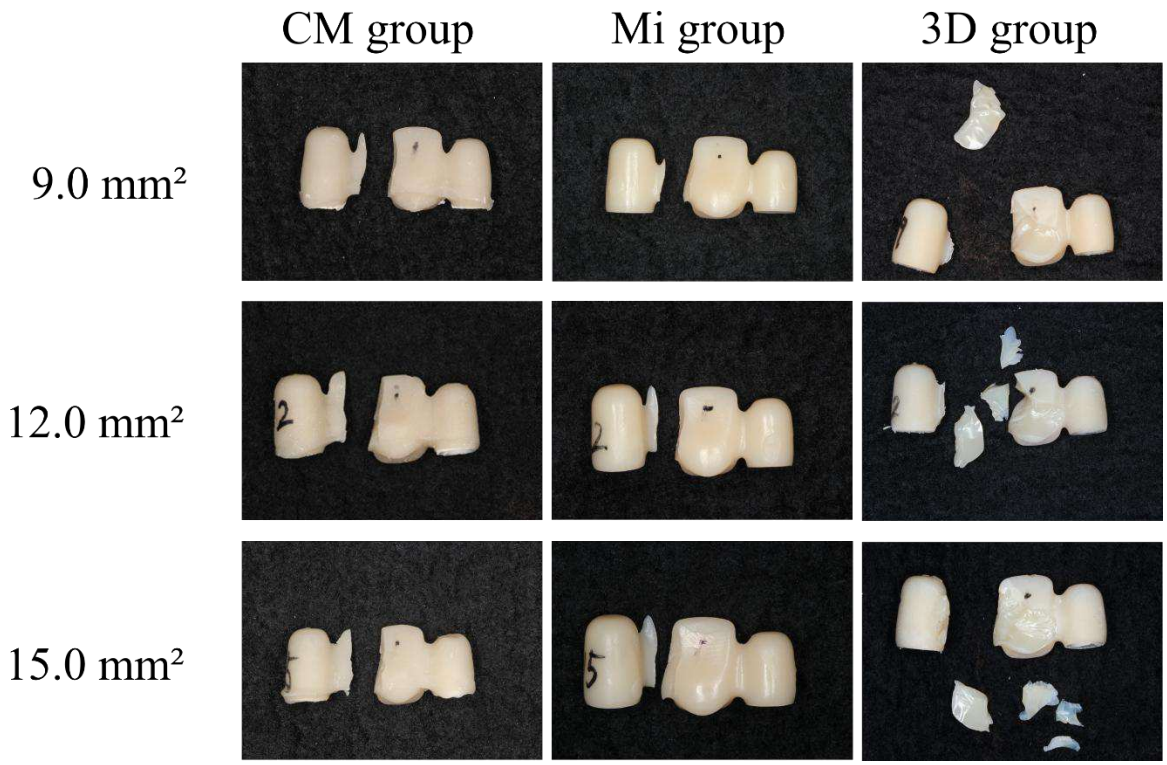

Mi group
Fracture load N
700
600
500
400
300
200
100
0
a
a
b
9.0mm²
12.0mm²
15.0mm²

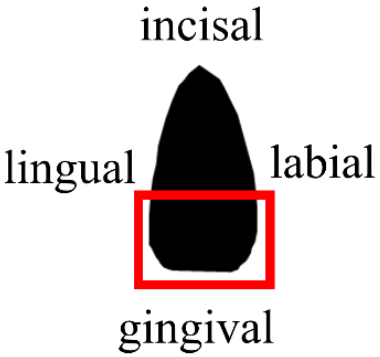

3D group
Fracture load N
700
600
500
400
300
200
100
0
a
b
c
9.0mm²
12.0mm²
15.0mm²

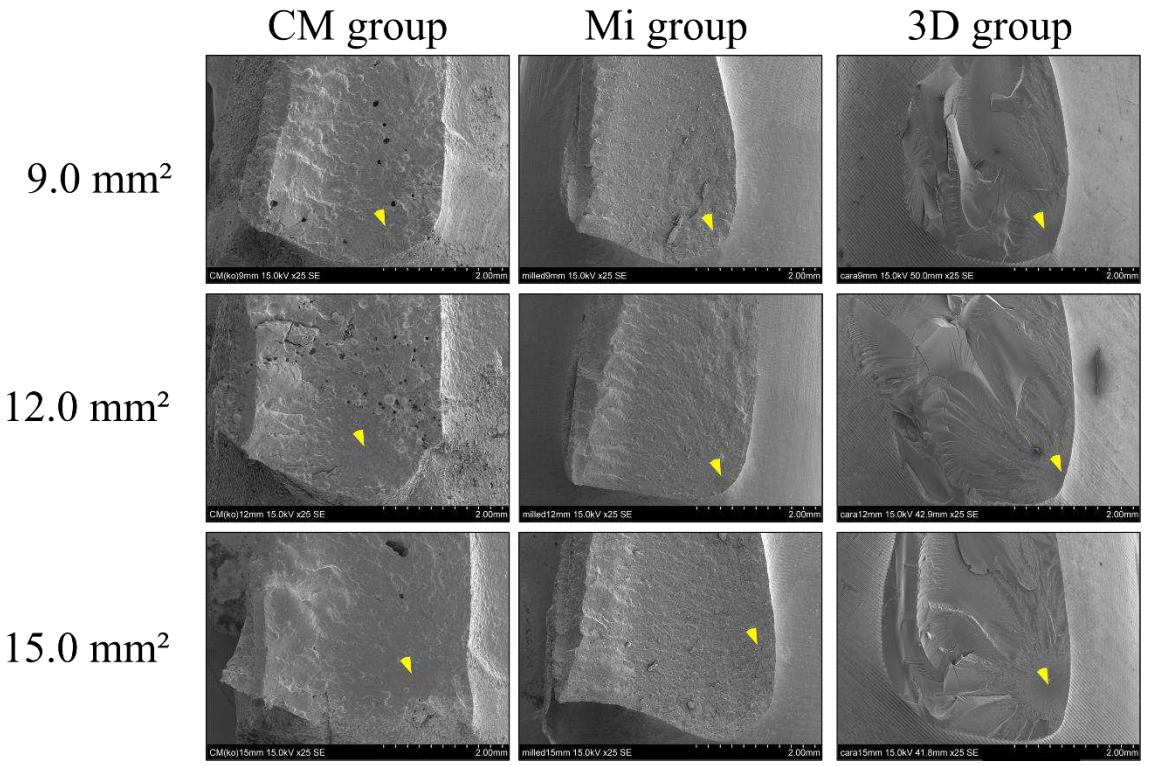

9.0mm²
Fracture load N
700
600
500
400
300
200
100
0
a
a
b
CM group
Mi group
3D group


12.0mm²
Fracture load N
700
600
500
400
300
200
100
0
a
a
b
CM group
Mi group
3D group


15.0mm²
Fracture load N
700
600
500
400
300
200
100
0
a
b
c
CM group
Mi group
3D group


CM group
Mi group
3D group
9.0 mm²
12.0 mm²
15.0 mm²


incisal
lingual
labial
gingival


CM group
Mi group
3D group
9.0 mm²
12.0 mm²
15.0 mm²