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A comparative evaluation of the mechanical and surface properties of reinforced 3D printed polymethylmethacrylate and CAD CAM polymethylmethacrylate - An in-vitro study

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Abstract

Introduction: Polymethylmethacrylate (PMMA) denture bases fabricated through 3D printing technology are proven to be better in comparison to its digital counterpart, the subtractive CAD CAM technique, in terms of cost effectiveness, reduced wastage, better detail reproduction and ability to print complex geometries. However, the mechanical and surface properties of 3D printed denture bases are found to be inferior to both conventional and CAD CAM techniques. Therefore, it is necessary to enhance the properties of 3D printed PMMA resin to make it a viable alternative to conventional and CAD CAM techniques.

Aim: To compare and evaluate the mechanical and surface properties of reinforced 3D printed PMMA with CAD CAM PMMA.

Method: A total of 72 samples of dimensions 64x10x3.3mm were divided into 4 groups (n=18), Group

I CAD CAM PMMA, Group II unmodified 3D printed PMMA, Group III 1wt% Al2O3 reinforced 3D printed PMMA and Group IV 2.5wt% ZrO2 reinforced 3D printed PMMA. Immediate flexural strength, surface hardness and surface roughness were tested after immersion in distilled water for 24 hours. Delayed flexural strength, surface hardness and surface roughness were tested after immersion in artificial saliva for 30 days followed by 5000 cycles of thermocycling simulating 6 months of intraoral use.

Results: The statistical analysis was performed using Kruskal-Wallis test and Wilcoxon signed-rank with significance set at p value < 0.05. Statistically significant differences ($p < 0.05$) were found among all groups. CAD CAM PMMA (Group I) had the highest immediate (116.20 ± 1.28 MPa) and delayed (108.25 ± 0.98 MPa) flexural strength, hardness (24.38 ± 0.64 VHN immediate; 22.82 ± 0.33 VHN delayed), and lowest roughness (0.66 ± 0.02 µm immediate; 0.74 ± 0.02 µm delayed). Zirconium oxide reinforcement (Group IV) showed higher flexural strength than Aluminum oxide (Group III), whereas Aluminum oxide reinforcement yielded the lowest surface roughness among the 3D printed groups.

Conclusion: Reinforcing 3D printed PMMA denture base resin with 2.5 wt% aluminum oxide or 1 wt% zirconium oxide significantly improved its mechanical and surface properties. While CAD-CAM PMMA remains superior, nanoparticle reinforcement offers a viable path to enhance the clinical performance of 3D printed resins.

Keywords: 3D printed PMMA, nanoparticles, Zirconium oxide, Aluminium oxide

Introduction

The increase in life expectancy has led to a rise in the geriatric population, with edentulism being a prevalent oral health problem.[1] Edentulous patients face functional, esthetic, and psychosocial challenges, and complete denture rehabilitation remains the most common treatment modality. The success of complete dentures depends on patient factors, the choice of material, and the fabrication method.[2]

Polymethylmethacrylate (PMMA), introduced by Walter Wright in 1936, has been the material of choice for denture base fabrication due to its favorable mechanical, physical, and esthetic properties. Conventional heat polymerization techniques such as compression molding and injection molding are widely used but have drawbacks such as porosity, polymerization shrinkage, surface roughness, and processing errors.[1,3]

Advancements in digital dentistry have introduced CAD CAM and 3D printing technologies for denture fabrication. CAD CAM milling uses pre-polymerized PMMA blanks fabricated under high temperature and pressure, providing superior mechanical and surface properties with minimal porosity and reduced microbial adhesion. However, the subtractive method involves significant material wastage and high fabrication costs.[4,5]

3D printing, an additive manufacturing technique, offers advantages such as reduced material waste, lower cost, faster production, and the ability to produce complex geometries. Nevertheless, 3D printed PMMA resins exhibit inferior mechanical and surface properties compared to conventional and CAD CAM PMMA. [4,5]

Reinforcement of heat cured PMMA with metal oxides nanoparticles has been shown to improve the mechanical, surface, and antimicrobial properties.[6]

Limited literature exists on nanoparticle reinforcement of 3D printed PMMA denture base resins. This study aims to evaluate and compare the immediate and delayed mechanical and surface properties of reinforced 3D printed PMMA with CAD CAM PMMA

Materials and Methodology

A total of 72 samples of dimensions 64x10x3.3mm were digitally designed and divided into 4 groups (n=18)

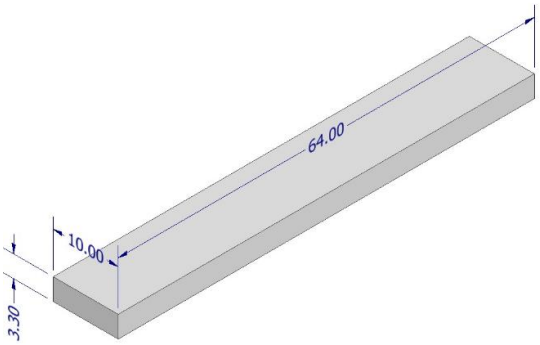

Figure 1: STL file of sample design

- **Group I:** CAD CAM PMMA
- **Group II:** Unmodified 3D printed PMMA
- **Group III:** 2.5 wt% Aluminum oxide–reinforced 3D printed PMMA
- **Group IV:** 1 wt% Zirconium oxide–reinforced 3D printed PMMA

Specimen Preparation

Group I- CAD CAM specimen preparation: 18 CADCAM specimen were designed in the dimensions of (64 × 10 × 3.3mm) and milled to be tested for flexural strength, surface hardness and surface roughness. The specimens were cleaned using isopropyl alcohol bath with 99% purity for 10 minutes.

Figure 2: CAD CAM specimen preparation

Group II- Unmodified 3D printed PMMA specimen preparation: 18 unmodified 3D printed PMMA specimen with dimensions of (64 × 10 × 3.3 mm) were printed at 90º orientation to be tested for flexural strength, surface hardness and surface roughness. The 3D printing process was followed by thorough washing of specimens in an isopropyl alcohol bath with 99% purity for 10 minutes.

Figure 3: Unmodified 3D printed PMMA specimen preparation

Group III- 2.5wt% Aluminium oxide nanoparticles reinforced 3D printed PMMA preparation: Nanocomposite was prepared by gradual incorporation of 2.5wt% of Al_3O_2 nanoparticles into the 3D printed PMMA solution under continuous magnetic stirring (Remi 1 MLH) for 30 minutes followed by mechanical mixing for 30 minutes at low frequency. The resultant resin was then used to print 18 aluminium oxide reinforced 3D printed PMMA specimen with dimensions of (64 × 10 × 3.3 mm) at 90º orientation to be tested for flexural strength, surface hardness and surface roughness. The 3D printing process was followed by thorough washing of specimens in an isopropyl alcohol bath with 99% purity for 10 minutes

Figure 4: 2.5wt% Aluminium oxide nanoparticles reinforced 3D printed PMMA preparation

GROUP IV- 1wt% Zirconium oxide nanoparticles reinforced 3D printed PMMA preparation: Nanocomposite was prepared by gradual incorporation of 1wt% of ZrO2 nanoparticles into the 3D printed PMMA solution under continuous magnetic stirring (Remi 1 MLH) for 30 minutes followed by mechanical mixing for 30 minutes at low frequency. The resultant resin was then used to print 18 zirconium oxide reinforced 3D printed PMMA specimen with dimensions of (64 × 10 × 3.3 mm) at 90o orientation to be tested for flexural strength, surface hardness and surface roughness. The 3D printing process was followed by thorough washing of specimens in an isopropyl alcohol bath with 99% purity for 10 minutes.

Figure 5: 1wt% Zirconium oxide nanoparticles reinforced 3D printed PMMA preparation

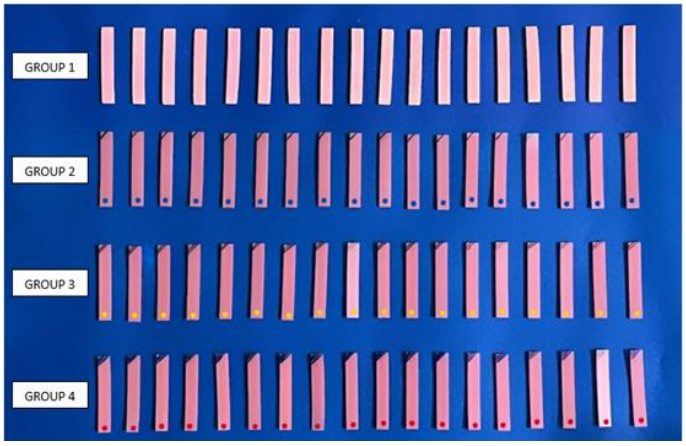

Figure 6: Samples categorized into respective groups

Artificial ageing: Twelve specimen from each group were stored in artificial saliva at 37ºC for 30 days and subjected to thermocycling for 5000 cycles between 5ºC to 55ºC with dwell time of 30 seconds each, equivalent to 6 months of intraoral exposure.

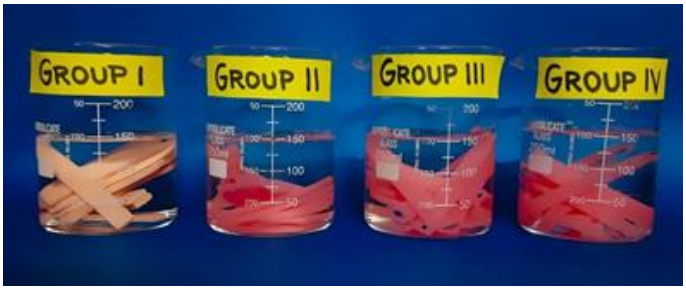

Figure 7: Storage in artificial saliva for delayed testing

Figure 8: Thermocycling of samples for delayed testing

Testing for flexural strength: A three-point bending test was used to evaluate the flexural strength. The load was applied at the centre of the specimen until fracture. The fracture load (N) was recorded to calculate the flexural strength (MPa) using the equation FS = 3Fl/2bh2, where F= fracture load, l= length of specimen, b= width of the specimen and h= thickness of the specimen

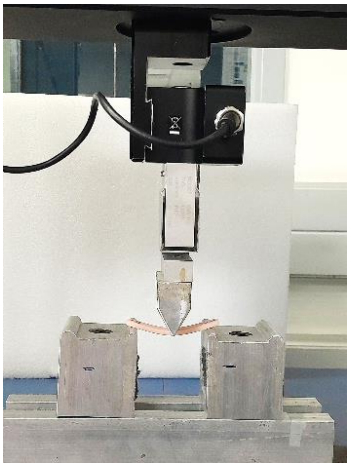

Figure 9: 3-point bending test for flexural strength using universal testing machine

Testing for surface hardness: Hardness was measured by Vickers hardness test and each specimen was subjected to a 50 N of load on three different sites. The final hardness (VHN) value of each specimen was arithmetically calculated by obtaining the average of the three readings.

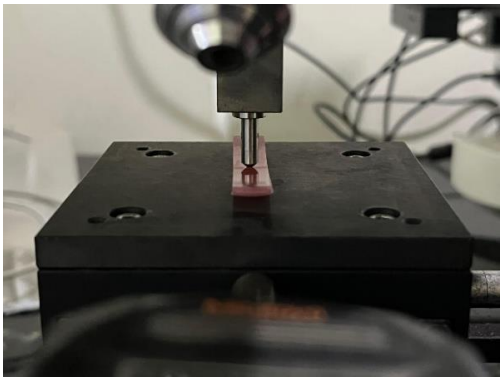

Figure 10: Vickers Hardness test for surface hardness

Testing for surface roughness: A non-contact optical profilometer was used for the measurement of the specimens' surface roughness. The specimens were radially scanned 3 times at different points with 0.01 mm resolution and the average surface roughness (µm) for each specimen was calculated.

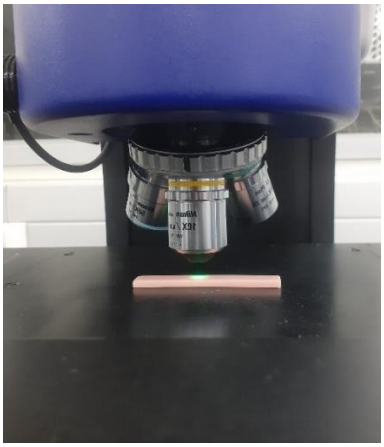

Figure 11: Non optical profilometer test for surface roughness

Statistical analysis: All statistical analyses were performed using IBM SPSS Statistics for Windows, Version 20.0 (IBM Corp., Armonk, NY, USA). Descriptive statistics were utilized to summarize quantitative data, with continuous variables expressed as means and standard deviations (SD) or medians with interquartile ranges (IQR), based on data distribution as assessed by the Shapiro-Wilk test. For intergroup comparisons of flexural strength, hardness, and roughness at both immediate and delayed time points, the Kruskal-Wallis test was employed owing to non-parametric data distribution, followed by pairwise comparisons using the Mann–Whitney U test with Bonferroni correction to control for type I error. The differences between immediate and delayed values within each group were analyzed using the Wilcoxon Signed-Rank test to determine the effect of artificial aging and thermocycling on material performance. A p-value of less than 0.05 was considered statistically significant.

Results

Table 1: Overall comparison of immediate flexural strength among the study groups

Groups	Mean±SD	Median (IQR)	Range	Chi-Square value; P value
Group 1	116.20±1.28	115.85 (115.13-117.40)	114.90-118.30	21.60; 0.001*
Group 2	86.33±1.22	86.30 (85.25-87.40)	84.80-88.00	
Group 3	98.63±0.99	98.45 (97.82-99.40)	97.60-100.30	
Group 4	104.17±0.93	104.25 (103.32-105.05)	102.80-105.20	

Kruskal Wallis test; *statistically significant ($p<0.05$)

Table 2: Overall Comparison of delayed flexural strength among the study groups

Groups	Mean±SD	Median (IQR)	Range	Chi-Square value; P value
Group 1	108.25±0.98	108.50 (107.30-109.12)	106.70-109.20	21.60; 0.001*
Group 2	76.80±0.89	76.75 (75.95-77.62)	75.80-78.00	
Group 3	91.77±0.42	91.80 (91.40-92.07)	91.10-92.30	
Group 4	98.35±1.01	98.60 (97.17-99.15)	97.10-99.60	

Kruskal Wallis test; *statistically significant ($p<0.05$)

Table 3: Comparison of immediate and delayed flexural strength

Groups	Immediate flexural strength	Delayed flexural strength	Z value; P value
Group 1	116.20±1.28	108.25±0.98	2.201; 0.028*
Group 2	86.33±1.22	76.80±0.89	2.207; 0.027*
Group 3	98.63±0.99	91.77±0.42	2.201; 0.028*
Group 4	104.17±0.93	98.35±1.01	2.207; 0.027*

Wilcoxon Signed Ranks test; *statistically significant ($p<0.05$)

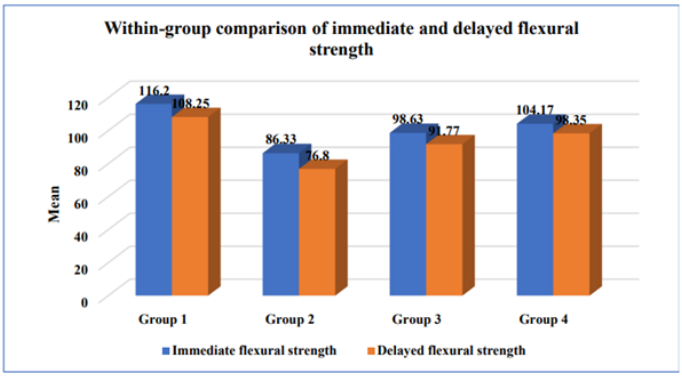

Figure 12: Within-group comparison of immediate and delayed flexural strength

CAD CAM PMMA (Group 1) showed the highest immediate flexural strength (116.20 ± 1.28 MPa), followed by ZrO_2 reinforced 3D-printed PMMA (Group 4, 104.17 ± 0.93 MPa), Al_2O_3 reinforced 3D-printed PMMA (Group 3, 98.63 ± 0.99 MPa), and unmodified 3D-printed PMMA (Group 2, 86.33 ± 1.22 MPa) as the lowest.

After aging, flexural strength decreased in all groups. CAD CAM PMMA (Group 1) remained highest (108.25 ± 0.98 MPa), followed by ZrO_2 reinforced (Group 4, 98.35 ± 1.01 MPa), Al_2O_3 reinforced (Group 3, 91.77 ± 0.42 MPa), and unmodified 3D-printed PMMA (Group 2, 76.80 ± 0.89 MPa).

Table 4: Overall comparison of immediate surface hardness among the study groups

Groups	Mean±SD	Median (IQR)	Range	Chi-Square value; P value
Group 1	24.38±0.64	24.40 (23.75-24.97)	23.60-25.20	20.95; 0.001*
Group 2	17.50±0.41	17.45 (17.20-17.87)	16.90-18.10	
Group 3	19.92±0.29	19.95 (19.65-20.15)	19.50-20.30	
Group 4	20.40±0.24	20.40 (20.17-20.62)	20.10-20.70	

Kruskal Wallis test; *statistically significant ($p<0.05$)

Table 5: Overall comparison of delayed surface hardness between the study groups

Groups	Mean±SD	Median (IQR)	Range	Chi-Square value; P value
Group 1	22.82±0.33	22.75 (22.50-23.15)	22.40-23.30	19.71; 0.001*
Group 2	16.15±0.19	16.15 (15.97-16.32)	15.90-16.40	
Group 3	18.95±0.33	18.85 (18.67-19.32)	18.60-19.40	
Group 4	19.08±0.15	19.05 (18.97-19.22)	18.90-19.30	

Kruskal Wallis test; *statistically significant (p<0.05)

Table 6: Comparison of immediate and delayed surface hardness

Groups	Immediate surface hardness	Delayed surface hardness	Z value; P value
Group 1	24.38±0.64	22.82±0.33	2.214; 0.027*
Group 2	17.50±0.41	16.15±0.19	2.201; 0.028*
Group 3	19.92±0.29	18.95±0.33	2.201; 0.028*
Group 4	20.40±0.24	19.08±0.15	2.201; 0.028*

Wilcoxon Signed Ranks test; *statistically significant (p<0.05)

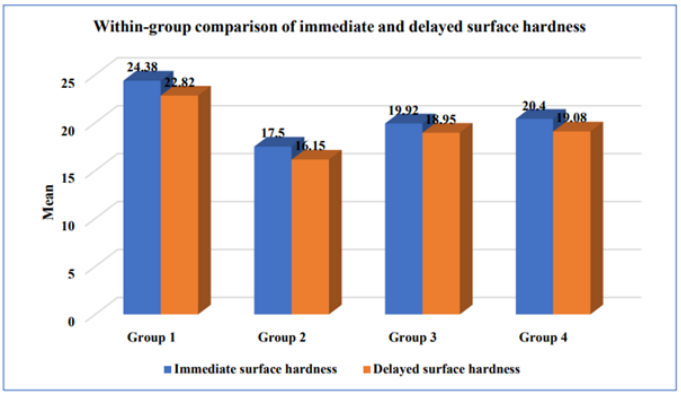

Figure 13: Within-group comparison of immediate and delayed surface hardness

For immediate hardness, CAD CAM PMMA (Group 1) recorded the highest mean value (24.38 ± 0.64 VHN), followed by ZrO_2 reinforced 3D-printed PMMA (Group 4, 20.40 ± 0.24 VHN), Al_2O_3 reinforced PMMA (Group 3, 19.92 ± 0.29 VHN), and unmodified 3D-printed PMMA (Group 2, 17.50 ± 0.41 VHN) as the lowest.

After thermocycling, all groups showed a decrease in hardness. CAD CAM PMMA (Group 1) remained hardest (22.82 ± 0.33 VHN), followed by ZrO_2 reinforced (Group 4, 19.08 ± 0.41 VHN), Al_2O_3 reinforced (Group 3, 18.95 ± 0.25 VHN), and unmodified 3D-printed PMMA (Group 2, 16.15 ± 0.19 VHN).

Table 7: Overall Comparison of immediate surface roughness among the study groups

Groups	Mean±SD	Median (IQR)	Range	Chi-Square value; P value
Group 1	0.66±0.02	0.65 (0.64-0.67)	0.63-0.68	16.28; 0.001*
Group 2	0.92±0.01	0.91 (0.91-0.93)	0.90-0.94	
Group 3	0.91±0.02	0.91 (0.88-0.92)	0.88-0.93	
Group 4	0.93±0.01	0.93 (0.92-0.94)	0.91-0.95	

Kruskal Wallis test; *statistically significant (p<0.05)

Table 8: Overall comparison of delayed surface roughness among the study groups

Groups	Mean±SD	Median (IQR)	Range	Chi-Square value; P value
Group 1	0.74±0.02	0.73 (0.72-0.75)	0.71-0.76	20.96; 0.001*
Group 2	1.06±0.02	1.05 (1.03-1.07)	1.03-1.08	
Group 3	0.94±0.02	0.94 (0.91-0.95)	0.91-0.96	
Group 4	1.01±0.03	1.01 (0.98-1.03)	0.97-1.05	

Kruskal Wallis test; *statistically significant (p<0.05)

Table 9: Comparison of immediate and delayed surface roughness

Groups	Immediate surface roughness	Delayed surface roughness	P value
Group 1	0.66±0.02	0.74±0.02	2.207; 0.027*
Group 2	0.92±0.01	1.06±0.02	2.220; 0.026*
Group 3	0.91±0.02	0.94±0.02	1.841; 0.066
Group 4	0.93±0.01	1.01±0.03	2.214; 0.027*

Wilcoxon Signed Ranks test; *statistically significant ($p<0.05$)

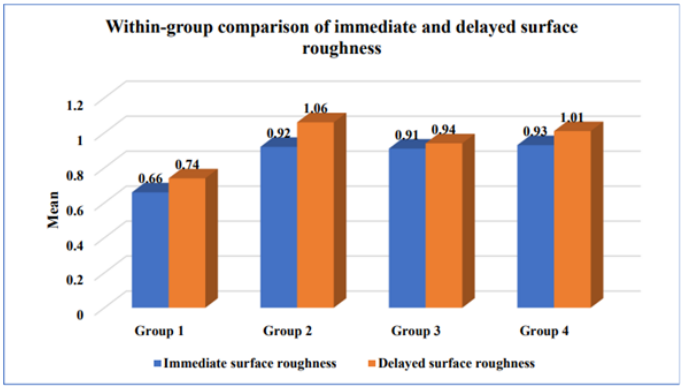

Figure 14: Within-group comparison of immediate and delayed surface hardness

Immediately after fabrication, CAD CAM PMMA (Group 1) showed the least surface roughness (0.66 ± 0.02 µm), followed by Al_2O_3 reinforced 3D-printed PMMA (Group 3, 0.91 ± 0.02 µm), ZrO_2 reinforced (Group 4, 0.93 ± 0.01 µm), and unmodified 3D-printed PMMA (Group 2, 0.92 ± 0.01 µm) exhibited the highest surface roughness.

After aging, surface roughness increased for all groups. However, CAD CAM PMMA (Group 1) still exhibited the lowest surface roughness (0.74 ± 0.02 µm), followed by Al_2O_3 reinforced PMMA (Group 3, 0.94 ± 0.02 µm), ZrO_2 reinforced PMMA (Group 4, 1.01 ± 0.03 µm), and unmodified 3D-printed PMMA (Group 2, 1.06 ± 0.02 µm) showed the highest roughness.

Discussion

The denture bases fabricated through 3D printing technology have proven to be better in comparison to their digital counterpart, the subtractive CAD CAM technique, in terms of cost effectiveness, reduced wastage, better detail reproduction and ability to print complex geometries.[5,6] However, the mechanical and surface properties of 3D printed denture bases are found to be inferior to both conventional and CAD CAM techniques.[1,7,8]

Several studies aimed at enhancing the properties of heat cured PMMA have stated that reinforcing PMMA with various fibres, and nanoparticles have significantly improved their physical, mechanical and surface properties. The most commonly used fibres for reinforcement are glass fibres, polyamide, polyethylene and polypropylene and natural fibres. The fillers included aluminium oxide, zirconium oxide, titanium oxide, silver, gold, platinum, palladium, silicon dioxide, hydroxyapatite, mica, and carbon in micro and nanosizes.[9,10, 11, 12, 13, 14]

This in-vitro study evaluated the flexural strength, surface hardness and surface roughness of unmodified 3D printed denture base PMMA resin and 2.5wt% Aluminium oxide and 1wt% Zirconium oxide nanoparticle reinforced 3D printed denture base PMMA resin and compared them with CAD CAM denture base PMMA resin. The specimens were tested for immediate mechanical and surface properties after 24-hour immersion in distilled water. To evaluate the long-term performance, the specimens were also tested for delayed properties after undergoing artificial aging to. To simulate the dynamic intraoral conditions, the specimens were immersed in artificial saliva for 30 days, followed by 5000 thermocycling cycles, which approximates six months of intraoral use.

Flexural strength represents the maximum stress a material can withstand before fracture under bending and is critical for denture durability. Consistent with earlier findings by Prpic et al[7]. and Gad et al.[15], CAD CAM PMMA showed the highest strength due to dense pre-polymerization under high temperature and pressure, producing a cross-linked, low-porosity structure. The unmodified 3D printed PMMA recorded the lowest strength, attributable to incomplete polymerization and weak interlayer adhesion during additive fabrication.

Layer thickness, printing orientation, and post-curing directly influence mechanical outcomes, with thinner layers enhancing dimensional stability and polymerization.[8,15]

Reinforced 3D printed groups demonstrated improved flexural strength over unmodified PMMA. Al_2O_3 reinforcement (2.5 wt%) significantly increased strength, consistent with Vojdani et al.[16], who attributed the improvement to crack-deflection mechanisms of alumina fillers. SEM analysis in previous studies confirmed uniform Al_2O_3 dispersion with minimal void formation at optimal concentrations, while excessive filler load (> 5 wt%) reduced strength due to agglomeration. Similarly, ZrO_2 reinforced PMMA (1 wt%) exhibited higher strength than both unmodified and Al_2O_3 reinforced resins, aligning with Alshaikh et al.[1] and Azmy et al[17]. This can be explained by ZrO_2's transformation from tetragonal to monoclinic phase, which absorbs crack-propagation energy and improves fracture resistance.

In the present study, the ZrO_2 group surpassed the Al_2O_3 group for both immediate and delayed testing, demonstrating superior crack deflection and stability. Nevertheless, CAD CAM PMMA remained superior overall due to its pre-polymerized nature. Thermocycling decreased flexural strength in all groups owing to water-induced polymer degradation. The least deterioration occurred in the ZrO_2-reinforced group, followed by Al_2O_3-reinforced, CAD CAM, and unmodified PMMA, indicating that nanoparticle reinforcement enhances resistance to thermal fatigue.

Surface hardness determines a material's resistance to wear and deformation. Low hardness can lead to surface abrasion, plaque retention, and color alteration, compromising longevity.[1] CAD CAM and heat-cured PMMA generally exhibit higher hardness than 3D printed PMMA, which has been attributed to material composition, water sorption, layer thickness, and printing orientation.[1,15]

Incorporation of nanoparticles substantially enhanced hardness. In this study, CAD CAM PMMA showed the highest hardness, followed by ZrO_2 reinforced, Al_2O_3 reinforced, and unmodified 3D printed PMMA. The difference between ZrO_2 and Al_2O_3 groups was statistically significant. The intrinsic hardness of Al_2O_3 is ≈15 GPa, which is lower than ZrO_2 which has a hardness of ≈20 GPa. This could explain the better results obtained by ZrO_2 in comparison to Al_2O_3. However, in agreement with previous studies,[9,10,17,18] the results confirm that both nanoparticles positively contribute to the improved surface hardness of 3D printed PMMA resin

Thermocycling significantly reduced surface hardness in all groups due to water absorption acting as a plasticizer within the PMMA matrix.[1,15] Among reinforced groups, Al_2O_3 reinforced PMMA demonstrated better retention of hardness than ZrO_2 reinforced PMMA after aging. The improved stability of Al_2O_3 may be attributed to its homogeneous dispersion and strong filler–matrix adhesion, which reduce filler debonding under stress. In contrast, phase transformation of ZrO_2 during thermal cycling can induce microstructural alterations that slightly compromise surface hardness

Surface roughness directly influences plaque accumulation, microbial colonization, and mucosal irritation. Highly polished denture surfaces are clinically desirable, as values above 2.2µm promote bacterial retention.[9] 3D printed PMMA typically shows greater surface roughness than CAD CAM PMMA due to its layer-by-layer structure and potential for incomplete interlayer fusion. Factors such as printing orientation, polishing technique, and resin composition also affect the final surface texture.[1,5,9,15]

In the present study, CAD CAM PMMA displayed the lowest roughness, followed by Al_2O_3 reinforced, unmodified, and ZrO -reinforced PMMA. The higher filler content and spherical morphology of Al_2O_3 particles enhanced packing density and polishability, yielding smoother surfaces than ZrO_2. Conversely, ZrO_2's angular particles resulted in slightly rougher surfaces.[1,16]. Several authors[1,9,17,19] reported minimal or no increase in roughness at low nanoparticle concentrations.

After thermocycling, roughness increased in all groups due to water absorption and thermal expansion, which weaken polymer chain bonds and induce microcracks. Reinforced groups exhibited less degradation than unmodified PMMA, confirming the stabilizing effect of nanoparticles. Al_2O_3 reinforcement showed the smallest increase (ΔRa = 0.03 μm), whereas ZrO_2 displayed greater change (ΔRa = 0.08 μm). The finer particle size and better interfacial bonding of Al_2O_3 minimize filler debonding and matrix disruption, accounting for its smoother surface after aging.[8,15]

The present study has inherent limitations that must be acknowledged. The study did not replicate the anatomical configuration of a denture. The stress distribution of a denture in its anatomical form would be much more complex than that of the rectangular specimens tested in this study. Also, the specimens were only subjected to static forces; the dynamic loading that the dentures must withstand during intraoral use was not replicated in this in vitro study. The artificial aging only simulated 6 months of intraoral use; further studies simulating longer durations of intraoral use are recommended. The 3D printing of the test specimen was carried out in only one configuration; further studies with different printing orientations and layer thickness are recommended. The current study only evaluated the effect of Al_2O_3 and ZrO_2; the effect of other nanoparticles with varying concentrations and the comparison between different nanoparticles should also be evaluated. Future studies should aim to overcome the aforementioned limitations to enhance the clinical relevance and applicability of the findings.

Clinical Implication

Reinforcing 3D printed PMMA with zirconium oxide or aluminum oxide nanoparticles significantly increases its mechanical and surface properties, bringing its performance closer to CAD CAM PMMA. This suggests that modified 3D printed resins can be used as a potential alternative for definitive denture bases, offering improved durability and reduced fabrication cost.

Conclusion

Within the limitations of this in-vitro study, it was concluded that:

- CAD-CAM PMMA exhibited the highest flexural strength and surface hardness, and the lowest surface roughness. However, a significant reduction in all three properties was observed on artificial aging.
- Unmodified 3D printed PMMA showed the least favourable results with the lowest flexural strength and surface hardness and highest surface roughness on both immediate and delayed testing.
- Reinforcement of 3D printed PMMA with 2.5 wt% Aluminium oxide significantly enhanced the immediate mechanical and surface properties. On delayed testing, Al_2O_3 showed better resistance to deterioration of surface properties.
- Reinforcement of 3D printed PMMA with 1wt% Zirconium oxide significantly improved the immediate flexural strength, surface hardness and surface roughness. However, on delayed testing, significant reduction in both mechanical and surface properties were observed.
- CAD-CAM PMMA exhibited the highest mechanical and surface properties, while unmodified 3D printed

PMMA presented the lowest values. However, reinforcing 3D printed PMMA with zirconium oxide significantly improved the immediate and delayed flexural strength, while aluminium oxide reinforcement exhibited significantly lower surface roughness and is better suited to resist surface degradation caused by thermal stresses.

In conclusion, nanoparticle reinforcement has proven to be a promising strategy to improve the mechanical and surface properties of 3D printed PMMA. However, this study has revealed that the choice of nanoparticles should be guided by the property that needs to be enhanced, as no single nanoparticle is capable of imparting all the desired properties.

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