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# Mechanical properties of 3D printed denture base polymers

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## ABSTRACT

**Statement of problem.** Studies on the mechanical properties of 3-dimensionally (3D) printed denture base polymers with appropriate test methods are lacking.

**Purpose.** The purpose of this in vitro study was to determine the flexural strength ($\sigma_f$), elastic modulus (E), fracture toughness ($K_{IC}$), work of fracture ($\omega_e$), and Martens hardness (HM) of 3D printed denture base polymers and to compare them with an injection molded material.

**Material and methods.** Three resins for additive (Lucitone Digital Print, LDP; Flexcera Base, FCB and an experimental material, EXP) and 1 for injection molding (IvoBase Hybrid, IBH) fabrication were analyzed. Standardized specimens were fabricated, polished, tempered, and thermal cycled (5 °C to 55 °C) before testing for $\sigma_f$, E, $K_{IC}$, $\omega_e$, and HM. The results were explored by global analysis ($\alpha$=.05). The Kolmogorov-Smirnov test was used to test for data distribution. Parametric and nonparametric tests followed by pairwise comparison were applied to test for differences between groups.

**Results.** The σf, E, and HM of 3D printed polymers were significantly lower than those of the injection molded, both for the tempered and aged groups. The σf was lowest for FCB and highest for IBH in the tempered state. The $K_{IC}$ and $\omega_e$ of the tempered EXP and IBH groups were lower compared with those of FCB and LDP. After aging, EXP, FCB, and IBH presented $K_{IC}$ in the same range, but it was lower than for LDP. Compared with the printable polymers, the control group IBH was not affected by artificial aging.

**Conclusions.** The $\sigma_f$, E, and HM of printable polymers were lower than those of the control group, and specimens did not fracture in bend testing. In contrast, $K_{IC}$ and $\omega_e$ were the highest for a printable polymer. Therefore, tension tests should be considered when testing ductile materials. (J Prosthet Dent 2025;133:1361.e1-e8)

Denture bases play an important role in removable prosthetics and have been fabricated using traditional methods such as injection molding or digitally, either by subtractive or by additive manufacturing (AM). The main AM processes used for resin-based dental materials include stereolithography (SLA) and digital light processing (DLP). As per the International Organization for Standardization (ISO)/ASTM 52900 standard,[1] DLP and SLA operate on the principle of vat polymerization, wherein resin layers are solidified through ultraviolet projected light or laser exposure. In contrast, for fused filament fabrication, molten filament material is deposited onto a build platform in a layer-by-layer process.[2] The advantages of AM lie in the reduced fabrication time and minimal material waste. Unlike milled denture bases where material loss occurs during milling and bur wear is a concern, only the material for supporting structures is lost in the AM process. The bases are printed layer by layer from resins followed by postprocessing and polymerization. The challenges of AM fabrication include that the denture base mechanical properties are influenced by the resin composition, the printing orientation and parameters, and the postprocessing steps (cleaning and postpolymerization).[3–7] Research to investigate and compare the mechanical properties of AM denture bases with those

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## Clinical Implications

The mechanical properties of 3D printed denture base polymers were affected by aging and vary depending on the material used and differ from polymers fabricated using injection molding. The 3D printed polymers investigated in this study deflected substantially before fracture.

fabricated using established conventional methods is essential.

A comparison of heat-polymerized, milled, and 3-dimensionally (3D) printed denture base resins showed lower flexural strength values for the AM objects.[8] However, a more recent investigation reported higher elastic indentation moduli and flexural strength values for AM resins compared with milled or conventionally polymerized ones.[9] Although, the required physical properties of conventionally fabricated denture base materials, such as injection moldable resins, have been standardized in the ISO 20795–1 standard,[10] some flexible injection molded dental base materials investigated in previous studies have not met these requirements.[11] Furthermore standards for testing 3D printing resins for definitive denture bases are currently lacking. The impact of aging on the mechanical properties of AM denture base materials in contrast with their conventional heat-polymerized counterparts have been reported in previous investigations.[12–15]

Given the different composition and behavior of these materials compared with traditional brittle denture base polymers, appropriate standardized tests should be established. Adapting test methods and specimen geometries from standards in the nondental polymer field could be an option.[16,17] These tests should evaluate mechanical properties, biocompatibility, bonding characteristics, color stability, and dimensional stability, as well as elastic and plastic deformation behavior. To simulate the oral environment, artificial aging methods should be incorporated.[18–20]

The objective of this study was to evaluate the mechanical properties of 3 AM denture base materials compared with a conventional injection molded denture base resin. The impact of aging was also investigated. The null hypotheses were that the denture base polymer or aging process would not affect the examined mechanical properties.

## MATERIAL AND METHODS

A total of 240 bar-shaped specimens were fabricated by 3D printing from 2 commercially available resins (Lucitone Digital Print, Lot No unavailable; Dentsply Sirona) (LDP), (Flexcera Base, Lot No: 042201; Desktop Health) (FCB) and an experimental resin (EXP) (Lot No: YM2226; Ivoclar AG) and by an injection molding resin (IvoBase Hybrid, Lot No: YB389T; Ivoclar AG) (IBH) (Fig. 1). EXP was printed with a 3D printer (PrograPrint PR5; Ivoclar AG), cleaned for 4 minutes in ≥97% isopropanol (IPA) (PrograPrint Clean; Ivoclar AG), and polymerized (PrograPrint Cure; Ivoclar AG) for 4.5 minutes. The FCB and LDP specimens were printed (M1; Carbon). LDP specimens underwent 2 cleaning cycles with ≥97% IPA, for 2 and 1 minute, in an ultrasonic bath, followed by drying and polymerizing (inLab SpeedCure; Dentsply Sirona) for 10 minutes on each side. The FCB specimens were cleaned in 2 cycles with ≥97% IPA, for 4 and 1 minute using a magnetic stirrer, and then dried and polymerized using 2×4000 flashes (Otoflash G171; NK-Optik). The control group (CG) specimens IBH were fabricated by injection molding (IvoBase Injector; Ivoclar AG). The specimens were stored for 20 days at room temperature (23 °C) in a dark environment, subsequently polished (Abramin; Struers) to their final dimensions with silicon carbide papers ranging from P500 to P1200, and divided into tempered and aged groups (Fig. 1). The final specimen dimensions for flexural strength ($\sigma_f$) were 3.3 ±0.2×10.0 ±0.2×64.0 mm, for Martens hardness (HM) 5.0×10.0×25.0 mm,

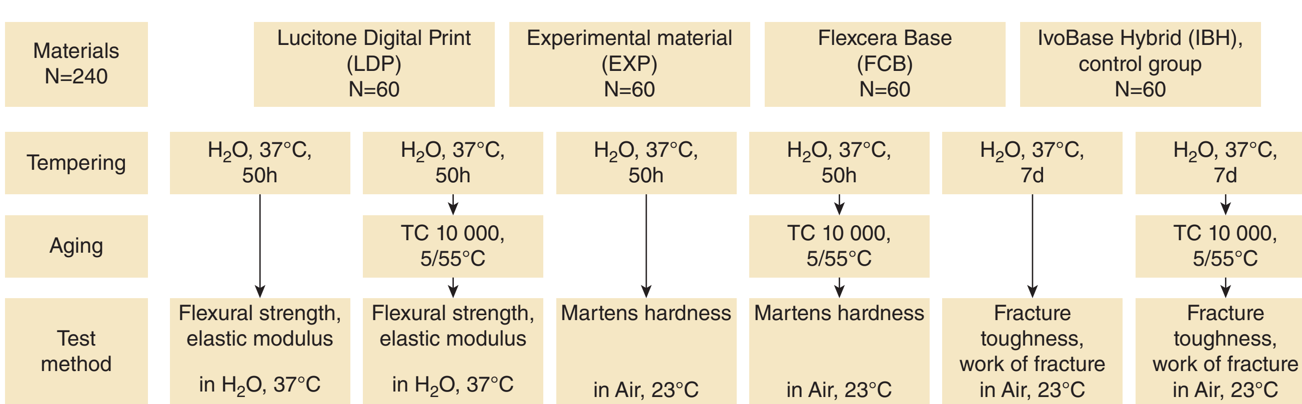


**Figure 1.** Study design.

and for fracture toughness ($K_{IC}$) 4.0 ±0.2×8.0 ±0.2×39.0 mm. In accordance with the ISO 20795–1 standard,[10] the specimens for $\sigma_f$ and HM testing were tempered (HERAcell 150; Thermo Fisher Scientific) for 50 hours, and the specimens for $K_{IC}$ measurements for 7 days, both in deionized water at 37 °C. Artificial aging (Thermocycler THE-1100; SD Mechatronik) was carried out by subjecting the specimens to 10 000 thermal cycles (5 °C and 55 °C), with a dwell time of 30 seconds at each temperature.

The 3-point bend $\sigma_f$ measurement was conducted by following the ISO 20795–1 standard[10] in a universal testing device (1445; Zwick/Roell) in 37 °C deionized water with a crosshead speed of 5 mm/minute in displacement-controlled mode until fracture or until 10 mm displacement. The ultimate $\sigma_f$ in MPa was calculated: $\sigma_f = 3Fl / 2wh^2$, where F=fracture load or maximum load in N; l=roller span (50 mm); and w and h=specimen width/height in mm. The elastic modulus (E) in GPa, was determined from the elastic region (0.05 to 0.25% strain range), of the stress-strain curves and was calculated from $E = [l^3(XH\text{-}XL)] / (4Lwh^3)$, where l=roller span in mm; XH and XL=end and start of elastic modulus determination in kN; L=bending between XH and XL; and w and h=specimen width and height in mm.

For HM, Vickers indentations with a force of 9.81 N and a dwell time of 10 seconds were applied using a hardness testing device (ZHU 0.2; Zwick/Roell) according to the ISO 14577–1 standard.[21] HM in MPa was determined from the force and indentation depth curve and was calculated from the equation $HM=F/A_S(h)=F/26.43h^2$, where HM=Martens hardness in MPa; F=test force in N; $A_S(h)$=surface area of the indenter penetrating beyond the zero-point of the contact, for Vickers diamond indenters $A_S(h)$=26.43; h=indentation depth in mm.

The $K_{IC}$ was analyzed according to the ISO 20795–1 standard.[10] A cutting machine (Secotom-50; Struers) with a 150-µm diamond cut-off wheel (M0D08; Struers) at a displacement speed of 0.05 mm/minute and rotational speed of 5000 rpm with water cooling was used to prepare the specimens. Subsequently, a sharp 0.1- to 0.4-mm notch was inserted at the bottom of the pre-notch using a custom-made device and razor blade. The dimensions of the cut (3.0 ±0.2 mm) and notch (0.1 to 0.4 mm) were verified microscopically (VHX-970F; Keyence) (Fig. 2). The $K_{IC}$ was measured in 3-point bending, with the notch positioned on the tension side and a crosshead speed of 1 mm/minute in a universal testing machine (1445; Zwick/Roell). Subsequently, the crack lengths were measured using a microscope (VHX-970F; Keyence) and $K_{IC}$ in $MPam^{1/2}$ was calculated: $K_{IC} = [(f\, P_{max}\, l_t) / (w_t\, h_t^{3/2})] \cdot \sqrt{10^{-3}}$, where $P_{max}$=fracture load in N; $l_t$: roller span (32 mm); $w_t$ and $h_t$=specimen thickness and height in mm; f=a geometric function of x: $f(x)=3x^{1/2} \cdot \{[1.99\text{-}x(1\text{-}x)(2.15–3.93x+2.7x^2)] / [2(1+2x)(1\text{-}x)^{3/2}]\}$ and $x=a / h_t$ with a=sum of the cut depth a′ (mm) and the notch depth (mm). The work of fracture, $\omega_e$ in J/$m^2$, was calculated by $\omega_e = \{U / [2w_t (h_t - a)]\} \cdot 1000$, according to ISO 20795–1 standard,[10] where U=area under the force displacement curve: $U = \int Pd\Delta$ in Nmm. $d\Delta$ is the corresponding deflection at load P.

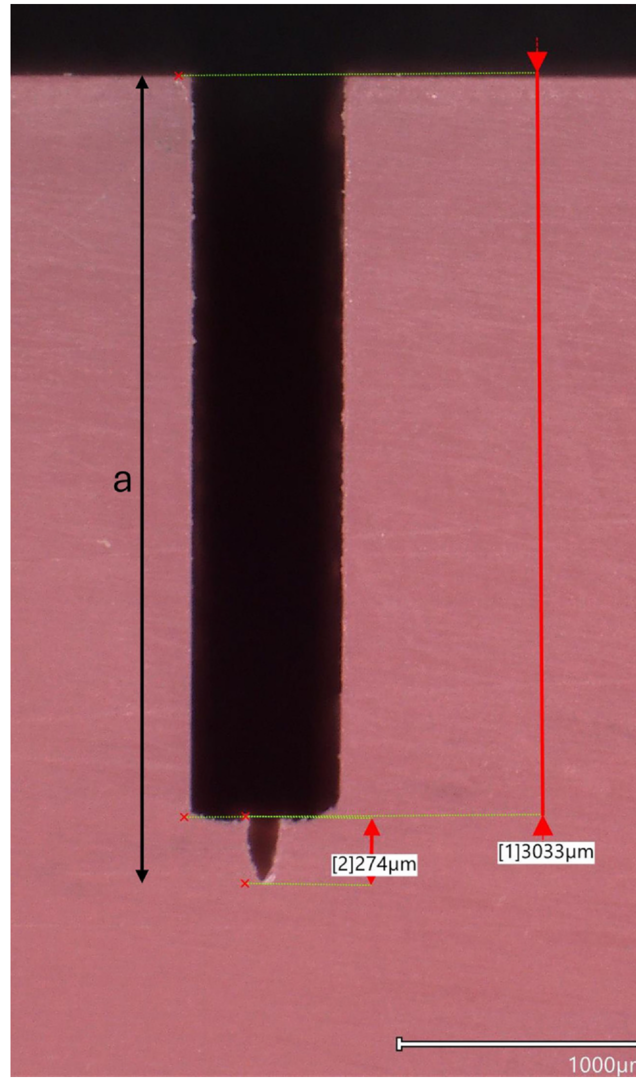

**Figure 2.** Representative SEVNB notch according to DIN EN ISO 20795–1 [ISO 20795–1:2013].

Statistical analysis was conducted using a statistical software program (IBM SPSS Statistics, v26; IBM Corp) ($\alpha$=.05). Descriptive statistics, including means, standard deviations (SD), and 95% confidence intervals (95% CI), were calculated. The normal distribution of the data was assessed using the Kolmogorov-Smirnov test. To evaluate the effect of parameters denture base resin material and aging on the $\sigma_f$, E, HM, $K_{IC,}$ and $\omega_e$, a multivariate 2-way ANOVA with post hoc Scheffé test and partial eta-squared ($\eta_p^2$) was performed. When the data did not meet normality assumptions and a significant interaction was observed, the data were divided by level of denture base resin materials and aging depending on the hypothesis of interest, and the Kruskal-Wallis test followed by pairwise comparison with Mann-Whitney-U-tests was used.

## RESULTS

The Kolmogorov-Smirnov test indicated a deviation from normal distribution for the $\sigma_f$, E, $K_{IC}$, and $\omega_e$ groups. Specifically, for each parameter, 1 of 8 groups (12.5%) exhibited nonnormal distribution, indicated by an asterisk in Tables 1 and 2. For $\sigma_f$, E, $K_{IC}$, and $\omega_e$, the 2-way ANOVA interaction (denture base resin materials versus aging) was significant ($P$<.001). The interaction between denture base resin materials and aging significantly affected the results ($P$<.001). Hence, the fixed effects of denture base resin materials and aging could not be compared directly as higher-order interactions were found to be significant. Consequently, nonparametric analyses were computed and divided by levels of denture base resin materials and aging depending on the hypothesis of interest. The HM data did not demonstrate a deviation from normal distribution or significant interaction between denture base resin materials and aging, which allowed for direct interpretation using the multivariate 2-way ANOVA with post hoc Scheffé test and partial eta-squared ($\eta_p^2$) (Tables 1, 3). The descriptive statistical results (Mean, SD, 95% CI) are presented in Tables 1 and 2 and as boxplots in Figure 3.

The stress strain curves (Fig. 4) indicated that IBH was the only material to fracture during the 3-point bend test, displaying the least strain, minimal plastic deformation, and the highest $\sigma_f$ (Fig. 3A). In contrast, all the printed denture base resin materials exhibited plastic deformation without fracture at a strain of 10 mm. LDP and FCB (both tempered and aged) reached an ultimate strength followed by a decrease in stress as strain increased. Aging had an impact on the $\sigma_f$ of LDP ($P$<.001), the EXP material ($P$<.001), and FCB ($P$<.001). Specifically, the $\sigma_f$ of LDP decreased for aged specimens from 52.6 to 39.3 MPa, while it increased for FCB aged specimens from 6.3 to 37.7 MPa. The $\sigma_f$ of the CG IBH was not significantly affected by aging ($P$=.055) and exhibited the highest $\sigma_f$ of 77.7 MPa (tempered) and 80.5 MPa (aged). The highest influence on the $\sigma_f$ (Fig. 3A) and E values (Fig. 3B) were the different denture base resin materials (partial eta-squared $\eta_p^2$>0.985, $P$<.001), followed by the interaction between denture base resin material and aging ($\eta_p^2$>0.918, $P$<.001) (Table 3). Aging had the lowest impact ($\eta_p^2$<0.299, $P$<.001) (Table 3). Within specimens tempered for 50 hours in distilled water at 37 °C, FCB showed the lowest $\sigma_f$ values of

**Table 1.** Descriptive statistics: mean ±standard deviation (SD) and 95% confidence intervals (95% CI) for ultimate flexural strength, elastic modulus and hardness values

| Aging | Denture Base Resin Material | $\sigma_f$, Ultimate Flexural Strength [MPa] | | E, Elastic Modulus [GPa] | | HM, Martens Hardness [MPa] | |
|---|---|---|---|---|---|---|---|
| | | Mean ±SD | 95% CI | Mean ±SD | 95% CI | Mean ±SD | 95% CI |
| tempered ($H_2O$,37 °C, 50 h) | LDP:Lucitone Digital Print | $52.6 \pm 1.6^{cB}$ | [50;54] | $1.84 \pm 0.2^{cB}$ | [1.6;2.0] | $127 \pm 13^{cA}$ | [116;138] |
| | EXP: Experimental material | $41.0 \pm 2.0^{bB}$ | [38;43] | $1.47 \pm 0.1^{bA}$ | [1.3;1.6] | $120 \pm 6^{bB}$ | [114;125] |
| | FCB:Flexcera Base | $6.3 \pm 1.7^{aA}$ | [4;8] | $0.32 \pm 0.1^{aA}$ | [0.1;0.4] | $57 \pm 12^{aA}$ | [55;75] |
| | IBH:IvoBase Hybrid | $77.7 \pm 3.3^{dA}$ | [74;81] | $2.70 \pm 0.1^{dA}$ | [2.5;2.8] | $168 \pm 8^{dA}$ | [161;174] |
| tempered ($H_2O$,37 °C, 50 h) + aged (TC 10 000, 5/55 °C) | LDP:Lucitone Digital Print | $39.3 \pm 2.7^{bA}$ | [36;42] | $1.44 \pm 0.1^{aA}$ | [1.2;1.6] | $123 \pm 8^{cA}$ | [116;130] |
| | EXP: Experimental material | $34.0 \pm 4.2^{aA}$ | [29;37] | $1.37 \pm 0.1^{aA}$ | [1.1;1.5] | $112 \pm 7^{bA}$ | [106;118] |
| | FCB:Flexcera Base | $37.7 \pm 2.1^{*abB}$ | [35;40] | $1.34 \pm 0.1^{aB}$ | [1.1;1.5] | $67 \pm 8^{aA}$ | [60;74] |
| | IBH:IvoBase Hybrid | $80.5 \pm 2.9^{cA}$ | [77;82] | $2.70 \pm 0.1^{*bA}$ | [2.5;2.8] | $169 \pm 5^{dA}$ | [164;173] |

Different lowercase letters indicate significant differences between tested denture base resin materials within one aging regimen
Different uppercase letters indicate significant differences between tested aging process within one denture base resin material
*deviation from normal distribution

**Table 2.** Descriptive statistics: mean ±standard deviation (SD) and 95% confidence intervals (95% CI) for fracture toughness and work of fracture values

| Aging | Denture Base Resin Material | $K_{IC}$, Fracture Toughness [MPam$^{1/2}$] | | $\omega_e$, Work of Fracture [J/m$^2$] | |
|---|---|---|---|---|---|
| | | Mean ±SD | 95% CI | Mean ±SD | 95% CI |
| tempered ($H_2O$,37 °C,7d) | LDP:Lucitone Digital Print | $2.29 \pm 0.2^{bA}$ | [2.0;2.5] | $1174 \pm 172^{bA}$ | [1040;1306] |
| | EXP:Experimental material | $1.23 \pm 0.1^{aA}$ | [1.0;1.3] | $239 \pm 52^{aA}$ | [200;276] |
| | FCB:Flexcera Base | $1.97 \pm 0.1^{bA}$ | [1.7;2.1] | $3741 \pm 301^{cA}$ | [3524;3957] |
| | IBH:IvoBase Hybrid | $1.28 \pm 0.1^{aA}$ | [1.1;1.4] | $182 \pm 15^{aA}$ | [170;193] |
| tempered ($H_2O$,37 °C,7d) + aged (TC 10 000, 5/55 °C) | LDP:Lucitone Digital Print | $2.24 \pm 0.1^{bA}$ | [2.0;2.4] | $1875 \pm 91^{bB}$ | [1808;1940] |
| | EXP:Experimental material | $1.11 \pm 0.1^{aB}$ | [0.9;1.3] | $264 \pm 76^{aA}$ | [208;319] |
| | FCB:Flexcera Base | $1.18 \pm 0.2^{aB}$ | [0.9;1.4] | $3317 \pm 736^{*cA}$ | [2789;3844] |
| | IBH:IvoBase Hybrid | $1.31 \pm 0.3^{*aA}$ | [1.0;1.6] | $196 \pm 40^{aA}$ | [166;225] |

Different lowercase letters indicate significant differences between tested denture base resin materials within one aging regimen
Different uppercase letters indicate significant differences between tested aging process within one denture base resin material
*deviation from normal distribution

**Table 3.** Two-way ANOVA results for comparison of all tested parameters with different denture base resin materials and aging

| Characteristics | Sum of Squares | *df* | Mean Squares | F | *P* | $\eta_p^2$ |
|---|---|---|---|---|---|---|
| **$\sigma_f$, Flexural Strength** | | | | | | |
| Constant parameters | 170563 | 1 | 170563 | 23117 | <.001 | 0.997 |
| Material | 34907 | 3 | 11636 | 1577 | <.001 | 0.985 |
| Aging | 227 | 1 | 227 | 31 | <.001 | 0.299 |
| Material*Aging | 5933 | 3 | 1978 | 268 | <.001 | 0.918 |
| Error | 531 | 72 | 7.4 | | | |
| Total | 212161 | 80 | | | | |
| E, Elastic modulus | | | | | | |
| Constant parameters | 218 | 1 | 218 | 20193 | <.001 | 0.996 |
| Material | 36 | 3 | 12 | 1120 | <.001 | 0.979 |
| Aging | .3 | 1 | .3 | 30 | <.001 | 0.294 |
| Material*Aging | 5.9 | 3 | 2 | 183 | <.001 | 0.884 |
| Error | .8 | 72 | .01 | | | |
| Total | 262 | 80 | | | | |
| HM, Martens hardness | | | | | | |
| Constant parameters | 1121294 | 1 | 1121294 | 14603 | <.001 | 0.995 |
| Material | 102360 | 3 | 34120 | 444 | <.001 | 0.949 |
| Aging | 103 | 1 | 103 | 1.3 | .251 | 0.018 |
| Material*Aging | 308 | 3 | 103 | 1.3 | .269 | 0.054 |
| Error | 5452 | 71 | 77 | | | |
| Total | 1243504 | 79 | | | | |
| $K_{IC}$, Fracture toughness | | | | | | |
| Constant parameters | 196 | 1 | 196 | 7721 | <.001 | 0.991 |
| Material | 14 | 3 | 4.6 | 181 | <.001 | 0.884 |
| Aging | 1.1 | 1 | 1.1 | 42 | <.001 | 0.371 |
| Material*Aging | 2.1 | 3 | .7 | 27 | <.001 | 0.536 |
| Error | 1.8 | 71 | .02 | | | |
| Total | 213 | 79 | | | | |
| $\omega_e$, work of fracture | | | | | | |
| Constant parameters | 148830451 | 1 | 148830451 | 1735 | <.001 | 0.961 |
| Material | 146595905 | 3 | 48865302 | 569 | <.001 | 0.960 |
| Aging | 123876 | 1 | 123876 | 1.4 | .234 | 0.020 |
| Material*Aging | 3132962 | 3 | 1044321 | 12.2 | <.001 | 0.340 |
| Error | 6091848 | 71 | 85801 | | | |
| Total | 305593850 | 79 | | | | |

6.3 MPa (*P*<.001) (Table 1), whereas the highest values were observed for IBH (77.7 MPa) (*P*<.001). After thermocycling, the EXP material showed a lower $\sigma_f$ (34.0 MPa) compared with LDP (39.3 MPa) and IBH (80.5 MPa) (*P*<.001). IBH presented the highest ultimate $\sigma_f$ (*P*<.001), and the EXP material was statistically similar to FCB (*P*=.078). Within LDP and the EXP material, the $\sigma_f$, E and HM values decreased after thermocycling (*P*<.001), except for E of the EXP material (*P*=.059). For FCB, an increase of values after thermocycling was observed (*P*<.001). Within IBH, no statistically significant impact of aging was found (*P*=.073).

The E values ranged from 0.32 to 2.7 GPa (Table 1, Fig. 3B). Within the HM values (ranging from 57 to 169 MPa) (Table 1, Fig. 3C), only the denture base resin material showed a statistically significant difference ($\eta_p^2$<0.949, *P*<.001) (Table 3).

The varying denture base resin materials had the highest influence on the $K_{IC}$ ($\eta_p^2$=0.884, *P*<.001), and aging had the lowest ($\eta_p^2$=0.371, *P*<.001) (Table 3). For tempered groups, EXP (1.23 $MPam^{1/2}$; 239 J/m²) and IBH (1.28 $MPam^{1/2}$; 182 J/m²) had lower $K_{IC}$ and $\omega_e$ values (*P*<.001) compared with FCB (1.97 $MPam^{1/2}$; 3741 J/m²) and LDP (2.29 $MPam^{1/2}$; 1174 J/m²) (Table 2, Fig. 3D). After aging, the $K_{IC}$ of EXP, FCB, and IBH were in the same range but were lower than LDP (*P*<.001) (Table 2, Fig. 3D). LDP (*P*=.400) and IBH (*P*=.165) were statistically similar after aging. FCB (*P*<.001) and EXP (*P*=.043) showed a decrease of $K_{IC}$ values after thermocycling. The $\omega_e$ of LDP increased after aging (*P*<.001).

## DISCUSSION

The objective of this study was to compare the mechanical properties of 3D printable resins for denture bases with conventional resins used in injection molding. The null hypotheses that the denture base polymer or aging process would not affect the mechanical properties were rejected.

The stress-strain curves presented in Figure 4 revealed that only the IBH (CG) bend bars fractured during 3-point bend $\sigma_f$ measurements, while the 3D printable denture base resin materials, LDP, the EXP material, and FCB, exhibited plastic deformation without recovery. The ultimate $\sigma_f$ at maximum load was lower than the $\sigma_f$ at a 10-mm bend for the 3D printable resins (Fig. 4). Therefore, the ultimate $\sigma_f$ was presented in Table 1, consistent with another study of highly flexible denture base resin materials.[11] At maximum load, the specimens experienced plastic deformation followed by the onset of relaxation and accompanied by a decrease in load. Because of bending relaxation, the strain increased while the load decreased, continuing either until fracture occurred or the set 10-mm deflection was reached.

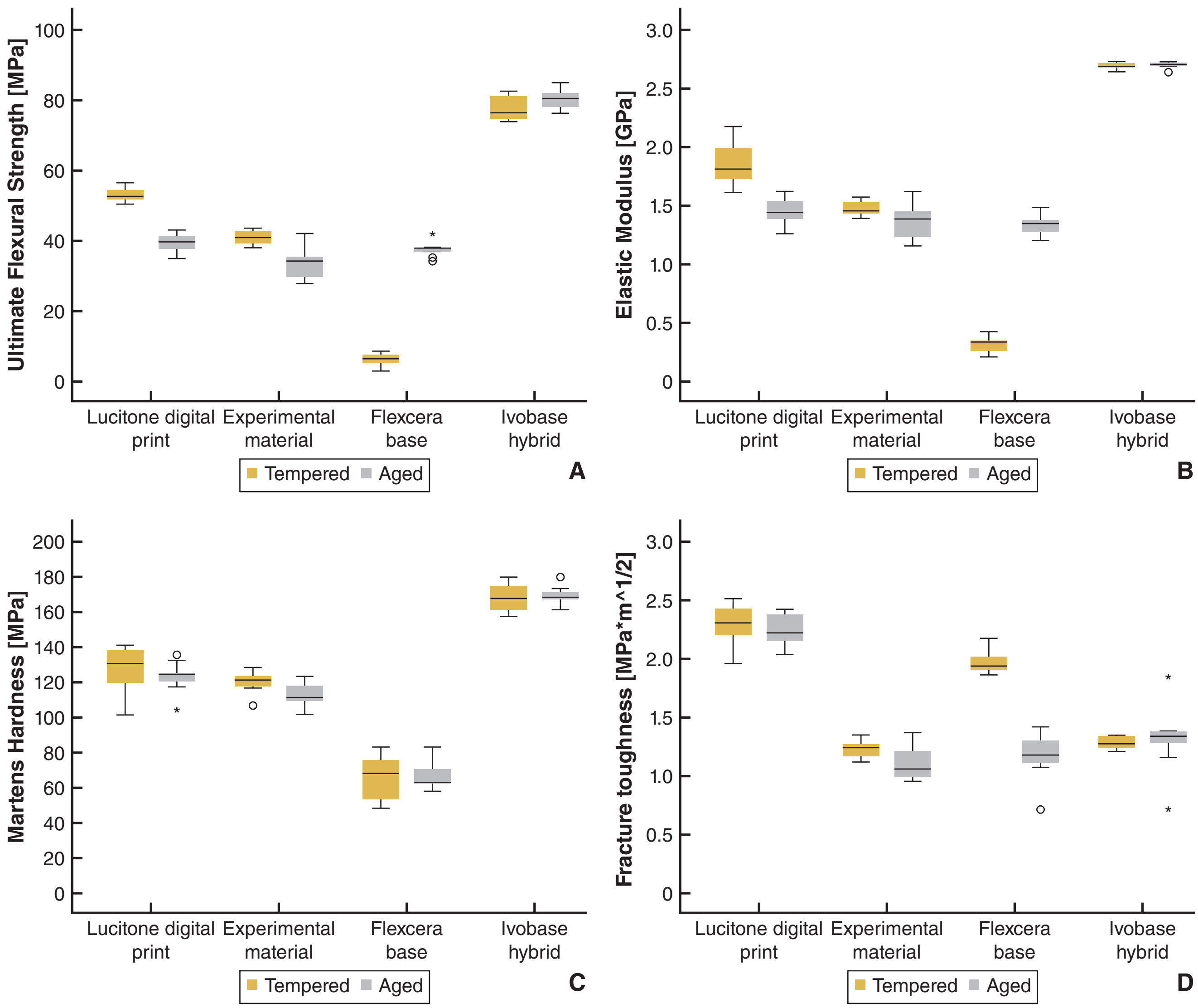


**Figure 3.** Boxplots showing different denture base resin materials, tempered (*white*) and aged specimens (*gray*). A, Ultimate flexural strength. B, Elastic modulus. C, Martens hardness. D, Fracture toughness. ○ represents outliers and * extreme outliers.

This is seen in Figure 4 for the materials LDP tempered and aged, the EXP material tempered, and FCB aged. In the case of IBH, the ultimate $\sigma_f$ corresponded to the fracture strength and was the only material to fracture during $\sigma_f$ testing. Unlike the tested printable resins, the $\sigma_f$ of the injection molded CG IBH remained unaffected by aging. These findings were consistent with those of previous studies[9,12–14] where thermocyclic aging caused a $\sigma_f$ decrease within 3D printed resins. The $\sigma_f$ of LDP and the EXP material decreased and that of FCB increased after thermocycling. In contrast with the present study, a previous study[15] reported that the $\sigma_f$ of Flexcera Base decreased after thermocycling. Because of thermocycling and the assumed postpolymerization of the FCB specimens, an increased $\sigma_f$ and decreased $K_{IC}$ were observed in the present study, indicating increased brittleness. Chemical processes such as water sorption, solubility, and the degree of conversion of residual monomers can occur during tempering and artificial aging, influencing the mechanical properties of polymer-based materials. The ISO standard 20795–1[10] requirement for the $\sigma_f$ of denture base polymers is 60 MPa. In contrast with a previous study,[22] the $\sigma$f values of the printable denture base resins in this study fell below the 60 MPa requirement. Measurements in the present study were conducted after storing the specimens for 20 days at room temperature in a dark environment, leading to a decrease in the $\sigma_f$ of approximately 15 MPa. The $\sigma_f$ divergence findings associated with postfabrication storage time was not part of the present study; however, it was additionally examined after comparing the $\sigma_f$ results of the present study with product specifications reported by manufacturers and with previous studies. The denture base polymer standard[10] for the $K_{IC}$ and $\omega_e$ states a minimum of 1.9 $MPam^{1/2}$ and 900 $J/m^2$ respectively. With the $K_{IC}$ and $\omega_e$ results of the present

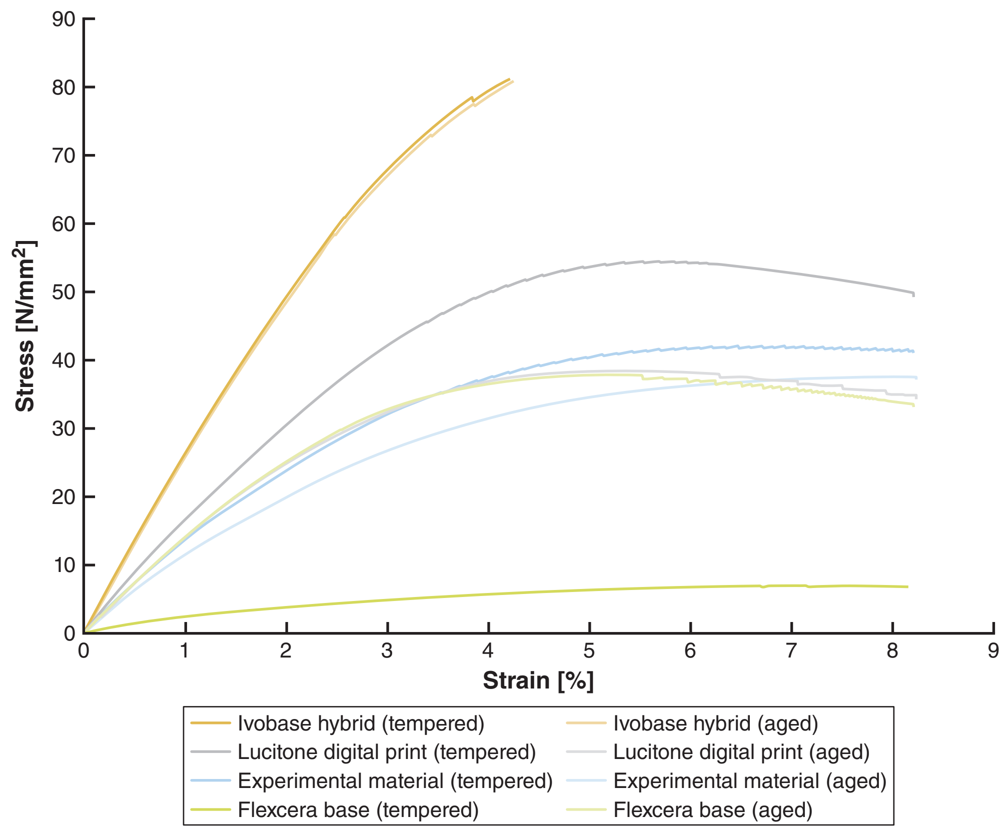


**Figure 4.** Representative stress-strain curves of materials during three-point-bending test.

study, the materials LDP ($K_{IC}$: 2.29 MPam$^{1/2}$; $\omega_e$: 1174 J/m$^2$) and FCB ($K_{IC}$: 1.97 MPam$^{1/2}$; $\omega_e$: 3741 J/m$^2$) are considered as high toughness, ductile materials for denture bases. These values were consistent with a mean 2.23 MPam$^{1/2}$ reported for the $K_{IC}$ of LDP in a previous study.[23] Thermocycling did affect the mechanical properties of 3D printed polymers but had no impact on the CG IBH. In contrast with the injection molded CG, the 3D printed bases with their interlayer areas[8] exhibited a reduced interlayer bond.[12,13]

Limitations of this study included that the specimens were stored for 20 days after fabrication before undergoing tempering, aging, and mechanical testing. The storage time of base polymers can impact their mechanical properties, and further studies should explore and vary the storage time to understand its effects better. Another limitation was the maximum displacement capability of 10 mm during the flexural strength and fracture toughness measurement associated with the test design. A higher displacement limit could have resulted in fractured bend bars, providing additional insights into the material behavior. Future investigations should consider higher deflection limits or alternative test designs such as tension tests to further evaluate the mechanical properties of denture base resin materials.

The use of 3D printable resins offers new opportunities for fabricating denture bases. However, the current ISO standard for testing base polymers primarily focuses on conventional, brittle polymers. Novel 3D printable resins have a wide range of elastic, elastic-plastic, and plastic deformation capabilities that may not be fully captured by the standard flexural strength and fracture toughness measurement methods involving bend tests. For ductile polymers with increased toughness, alternative strength test methods such as tension tests[16] with dumbbell shaped specimens[17] should be considered and incorporated into dental material testing standards. It is important for dental materials standards to address the characteristics of novel materials like printable resins. Additionally, although high bending forces may be clinically relevant for denture bases, sudden high-force impacts during mastication could have greater clinical significance. Future studies could consider simulating load impact in vitro during bend testing by increasing the crosshead speed, for example, to 20 mm/minute. This approach could provide a more realistic assessment of how these materials perform under dynamic loading conditions.

The 3D printed polymers examined in this study exhibited lower $\sigma_f$, E, and HM but increased $K_{IC}$ and $\omega_e$ compared with injection molding polymers. The lower E and higher $K_{IC}$ of 3D printed polymers may help prevent catastrophic crack propagation in denture bases by deflection instead. However, the high plastic deformation observed in complete denture bases made from these materials could lead to a loss of retention over time, affecting occlusion. The response of denture bases to applied loads, whether elastic or plastic, depends on factors such as mastication force and the clinical situation (complete or partial dentures). Unlike the injection molding polymer IBH, aging had an impact on the mechanical properties of the 3D printed resins. Understanding how these polymers behave in an oral environment under various loading conditions is crucial to selecting denture base resin materials based on specific clinical indications to ensure optimal performance and longevity in different patients. Further research is needed to investigate the behavior of these materials in realistic oral environments and loading situations.

## CONCLUSIONS

Based on the findings of this in vitro study, the following conclusions were drawn:

1. The CG material IBH exhibited the lowest strain and plastic deformation and highest $\sigma_f$, E, and HM.
2. The $K_{IC}$ and $\omega_e$ of IBH and the EXP material were lower than those of LDP and FCB.
3. Aging affected the mechanical properties of 3D printed resins but had no impact on the injection moldable IBH.
4. LDP and FCB can be considered as high toughness materials exceeding the ISO 20795-1:2013 standard limit of 1.9 $MPam^{1/2}$ $K_{IC}$ and 900 $J/m^2$ $\omega_e$.
5. The 3D printable polymers exhibited high bending without fractures during $\sigma_f$ testing, with plastic deformation and $\sigma_f$ values below the 60 MPa denture base standard. Therefore, a tension test design should be considered for highly bendable denture base resin materials.

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