Biomechanical Behavior of a 3D-Printed Denture Base Material

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Purpose: To evaluate relevant material properties (flexural strength [σ_f], elastic modulus [*E*], water sorption [Wsp] and solubility [Wsl], and biocompatibility) of an additive manufacturing (AM) polymer vs a heat-curing acrylic resin (AR; control) for the manufacture of complete dentures, testing the hypothesis that fabrications from both materials would present acceptable material properties for clinical use. ***Materials and Methods:*** The σ_f, *E*, Wsp, and Wsl were evaluated according to the ISO 20795-1:2013 standard, and the biocompatibility was evaluated using MTT and SRB assays. Disk-shaped specimens were fabricated and used for Wsp (n = 5), Wsl (n = 5), and biocompatibility (n = 3) testing. For assessment of σ_f and *E*, bar-shaped specimens (n = 30) were fabricated and stored in 37°C distilled water for 48 hours or 6 months before flexural testing in a universal testing machine with a constant displacement rate (5 ± 1 mm/minute). Data from σ_f, *E*, Wsp, Wsl, and biocompatibility tests were statistically analyzed using Student *t* test (α = .05). Weibull analysis was also used for σ_f and *E* data. ***Results:*** Significant differences between the two materials were found for the evaluated material properties. Water storage for 6 months did not affect the flexural strength of the AM polymer, but this material showed inadequate σ_f and Wsl values. ***Conclusions:*** Despite adequate biocompatibility and strength stability after 6 months of water storage, the AM polymer recommended for complete dentures needs further development to improve the material properties evaluated in this study. *Int J Prosthodont 2024;37(suppl):s109–s117. doi: 10.11607/ijp.8295*

Edentulism is a highly prevalent condition worldwide that is more prominent among low- and middle-income countries.[1] Tooth loss can negatively impact quality of life, decreasing functional and nutritional abilities,[2,3] affecting esthetics and lifestyle, and reducing self-esteem and social integration.[4,5] A complete denture is a treatment option to minimize such negative effects for edentulous patients.[6]

The American Society for Testing and Materials (ASTM) defines additive manufacturing (AM) as the process of joining materials to make objects from 3D model data, usually layer upon layer, as opposed to subtractive manufacturing methodologies.[7] Digital technologies have advanced rapidly in dentistry, enabling the development of faster and less expensive techniques with greater predictability,[7] including the manufacture of complete dentures and their parts.[8–10] As a result, AM technology is now able to fabricate structures from digital models created by computer-aided design (CAD) software and transferred to 3D printers. The transition to the clinical application of AM in dentistry is highly dependent on the available materials, which must not only provide the necessary precision but also the biologic and physical properties suitable for dental use.[11] Water sorption and solubility, flexural strength, elastic modulus, and biocompatibility are clinically relevant material properties for a denture base.[11–13] Acrylic resins (ARs), which are traditionally used to manufacture complete

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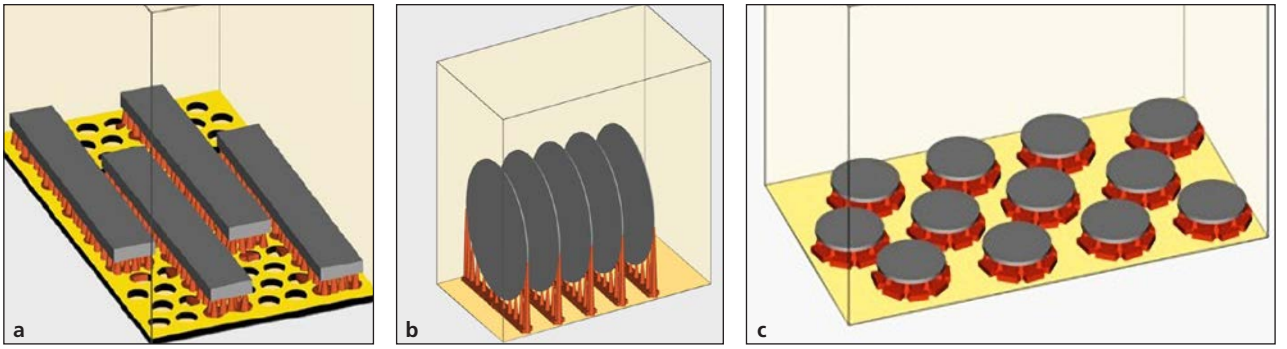

Fig 1 *(a)* Bar-shaped and *(b and c)* disk-shaped specimens were designed to fit on the digital printing platform.

denture bases, must withstand masticatory forces[12] and present adequate water sorption and solubility values, as these properties can influence the dimensional behavior and stability of the denture.[13]

Water can significantly affect the dimensional and mechanical properties of polymers, which absorb a small amount of water when in an aqueous system. Water penetrates the polymer chains, acting as a plasticizer and slightly expanding the polymerized material.[14] Flexural strength has been widely used to characterize materials and structures because it is an efficient and simple test.[15–17] Studies have evaluated the flexural strength (σ_f),[16,18] elastic modulus (*E*),[16,18] water sorption (Wsp),[19,20] and water solubility (Wsl)[19,20] of heat-curing ARs used for denture bases. However, there is limited information on the material properties of AM polymers used for denture bases.[21] The flexural strength of denture base resins manufactured with different techniques (conventional and AM) ranged from 62 to 110 MPa for conventional ARs, while an AM polymer (NextDent Base) ranged from 60 to 84 MPa.[22] The evaluation of postpolymerization time (0, 5, 10, and 20 minutes) on flexural strength, elastic modulus, fracture toughness, water sorption and solubility, degree of conversion, surface hardness, and cytotoxicity of resin materials for denture bases (3D-printing PMMA or conventional heat-curing resin) showed that all properties and parameters improved with increasing postcuring time for the AM resin, while 20 minutes of exposure had similar behavior for the conventional heat-curing resin.[23] In addition, other parameters influence the accuracy of printed objects, such as the printing technology, the type and color of resin, the type of polymerization light, and the procedures using for the postprinting process.[21]

Thus, AM technology could be used to improve the development of new and practical dental treatment options. However, knowledge about materials and the clinical applicability of AM in dentistry is still incipient. Therefore, the present study aimed to evaluate the biomechanical behavior and biocompatibility of an AM polymer used to fabricate complete denture bases by investigating clinically relevant material properties that were compared to a conventional AR also used for denture bases. The hypothesis tested was that both materials present acceptable properties for clinical use.

MATERIALS AND METHODS

Specimen Preparation

Disk-shaped specimens (50-mm diameter × 0.5-mm thickness for Wsp and Wsl tests; 4-mm diameter × 1-mm thickness for biocompatibility) and bar-shaped specimens (64-mm length × 10-mm width × 3.5-mm thickness for σ_f and *E* tests) were manufactured for the two materials,[24] AR (Clássico; control), and AM polymer (Cosmos Dental Model, Yller).

The 3D specimens were created and saved as standard tessellation language (STL) files that were transferred to the specific system (CAMcreator) of the 3D printer (Varseo, Bego) and organized three-dimensionally on the digital printing platform (Fig 1). The number and placement of the sprues followed the standard software suggestion.

Respecting the limit of the printing area of the 3D printer based on stereolithography (SLA) technology, each cycle allowed printing of four bar-shaped specimens or five disk-shaped specimens. The bar-shaped specimens were horizontally printed according to their largest surface and contained 182 layers with a thickness of 0.027 mm (27 µm) comprising the printing base, the support sprues, and the specimen. The disks were vertically printed and contained 1,080 layers with a thickness of 0.027 mm (27 µm) from the base to the specimen. After printing, the structure (including the printing base, sprues, and specimen) was removed from the printer, and the residue from the printing resin was initially removed in 96% ethyl alcohol. The specimens were then sonically cleaned in a digital ultrasonic cleaner (CD-4810, Gnatus) in a 96% ethyl alcohol bath for 10 minutes.

Postcuring was performed in a light-curing unit (Magnabox, EDG) consisting of a 3-minute cycle followed by two polymerization cycles of 5 minutes each (total time: 13 minutes). The specimens were manually separated from

the sprues, and small defects from removing the sprues were polished using a #2,500 sandpaper (3M ESPE) applied in one direction with water.

The bar- and disk-shaped AR specimens were conventionally fabricated using dental laboratory muffles and matrices made using printing resin (Cosmos Denture, Yller) and a 3D printer, as described previously. The lower part of the muffle was filled with a layer of extra hard plaster (Tuff Rock 44, Talladium). After setting, the plaster was covered with a layer of laboratory condensation silicone (Zettalabor, Zhermack). Four 3D-printed matrices were placed in the silicone, and a 2-kg glass plate was placed on them to obtain a smooth layer. After complete setting of the silicone, the middle part of the muffle was positioned, an isolation coat (Cel-Lac, SS White) was applied to the surface of the silicone/matrices, the muffle was filled with extra hard plaster, and the muffle was closed. After complete setting of the plaster, the muffle was opened and the matrices were removed, leaving their impression on the laboratory silicone. The heat-activated AR was manipulated according to the manufacturer's instructions and adapted to the impressions left by the matrices. The muffle was closed and hydraulically pressed to remove the excess AR. The muffle was placed in a manual press, immersed in water at room temperature, and heated to 70°C for 30 minutes, followed by 30 minutes of temperature maintenance and subsequent heating to 100°C for 60 minutes. The water was allowed to cool to room temperature, and the muffle was removed from the water and opened to access the AR specimens, which were carefully removed and polished in water with #2500 sandpaper.

The disk-shaped AR specimens (50 × 0.5 mm) for the Wsp and Wsl tests were obtained from a stainless steel matrix.[23] The AR was manipulated to the plastic phase as described previously and fitted in the metal matrix, which was closed with the stainless steel lid. The matrix was pressed (1,000 kg for 12 hours), the AR excess was removed, and the matrix was placed into a container with water at room temperature for the polymerization cycle as described previously. After AR polymerization and matrix cooling, the matrix was opened, and the specimens were removed.

All specimens were visually inspected to assess potential manufacturing failures.

Flexural Strength and Elastic Modulus

The bar-shaped AR and 3D specimens were tested for σ_f and *E* according to the ISO 20795-1:2013 standard.[24] After fabrication, all specimens were stored in 37°C distilled water for 48 hours or 6 months before submitting them to the 3-point bending test. The water of the specimens stored for 6 months was changed weekly, ensuring that the specimens remained constantly immersed in water.

The specimens (n = 30) were removed from the water storage and placed on the supporting rollers (3.2-mm diameter × 10.5-mm length; span = 50 mm) of the 3-point bending test unit immersed in 37°C water. The metallic piston of the universal testing machine (Instron) was carefully positioned in the center of the specimen, and the load was applied with a constant displacement rate of 5 mm/minute until specimen fracture.

Flexural strength was calculated (in MPa) with the following equation, where F is the maximum load (in Newtons), l is the span between the supports (50 mm), b is the width (in millimeters) of the specimen, and h is the height (in millimeters) of the specimen:

$$\sigma = 3Fl / 2bh^2$$

The *E* was calculated (in megapascals) with the following equation, where F1 is the load (in Newtons) at a point in the straight line portion (with maximum slope) of the load/deformation curve, d is deformation (in millimeters) at F1, and l, b, and h are as defined above:

$$E = F_1 l^3 / 4bh^3 d$$

The fractured surfaces of the specimens were evaluated based on fractography principles. Hence, a visual inspection was performed initially and followed by stereomicroscopy (Stemi 2000-C, Zeiss) and scanning electron microscopy (SEM; Vega3, Tescan) for representative specimens.

Water Sorption and Solubility

Disk-shaped specimens (n = 5) were used for Wsp and Wsl tests.[24] The specimens were placed in the rack inside the desiccator containing dried silica gel. The desiccator was placed in an oven at 37°C for 23 hours and then removed. The rack containing the specimens was transferred to a second desiccator with freshly dried silica gel that was kept at 23°C for 60 minutes before weighing the specimens with an analytical balance to an accuracy of 0.2 mg (Bioprecisa, TDS). The desiccator was kept sealed except for handling the specimens and changing the dry silica gel. After weighing the specimens, the silica gel in the desiccator was replaced with freshly dried gel, and the rack with the specimens was placed in the desiccator and then returned to the oven. This procedure was repeated until the specimens presented a constant mass, *m1* (conditioned mass); ie, until the loss in mass of each specimen was not more than 0.2 mg between successive weighings. At this point, the specimens were considered dry. The volume, *V* ($V = \pi r^2 h$), of each specimen was calculated using the mean of three diameter measurements and the mean of five thickness measurements.

The conditioned specimens were immersed in water[25] at 37°C for 7 days. They were then removed from the water, placed on a paper towel until they were free from visible moisture, waved in the air for 15 seconds, and

Table 1 Flexural Strength and Elastic Modulus of AR and AM Polymers According to Water Storage Time

Material	Property (MPa)	n	Time of assessment	
			48 h	6 mo
AR	σ_f	30	78.1 ± 5.5 (76.1–80.2)^{A,a}	52.8 ± 3.5 (51.4–54.1)^{B,a}
	E	30	1,707 ± 190 (1636–1778)^{A,a}	1,172 ± 99 (1134–1211)^{B,a}
AM polymer	σ_f	30	34.4 ± 7.5 (31.6–37.2)^{A,b}	31.9 ± 8.8 (28.6–35.2)^{A,b}
	E	30	811 ± 128 (759–862)^{A,b}	539 ± 115 (496–582)^{B,b}

Data are reported as mean ± SD (95% CI). Different uppercase letters indicate statistical differences in the same row ($P \leq 0.05$). Different lowercase letters indicate statistical differences in the same column for the same material property ($P \leq .05$).

Table 2 Weibull Modulus (m) and Characteristic Strength Values

Material	m, MPa	s_0, MPa
AR (at baseline)	17.5 (12.4, 22.3)	80.5 (78.7, 82.5)
AR (after 6 mo)	18.0 (12.3, 22.0)	54.3 (53.0, 55.7)
AM (at baseline)	5.5 (3.6, 6.5)	37.2 (34.1, 40.8)
AM (after 6 mo)	4.2 (3.0, 5.4)	35.0 (31.9, 38.7)

Data are presented as mean (95% CIs).

weighed 60 seconds after removal (to an accuracy of 0.2 mg). This mass was then recorded as *m2*.

The specimens were reconditioned to constant mass in the desiccator as described previously to obtain *m1*, and the mass of the "reconditioned" specimens was recorded as *m3*. The same conditions as in the first drying process for *m1* were applied.

The Wsp value for each specimen was expressed in micrograms per cubic millimeter (μg/mm^3) using the following equation, where m2 is the mass of the specimen (in micrograms) after water immersion, *m3* is the reconditioned mass of the specimen (in micrograms), and *V* is the volume of the specimen (in millimeters cubed):

$$Wsp = m2 - m3 / V$$

The soluble matter per unit volume (ie, Wsl) leached out during immersion, expressed in micrograms per cubic millimeter, was calculated for each specimen from the following equation, where *m1* is the conditioned mass of the specimen (in micrograms), and *m3* and *V* are as described above:

$$Wsl = m1 - m3 / V$$

The values calculated for Wsl were rounded off to the nearest of 0.1 μg/mm^3.

Biocompatibility

Primary gingival fibroblasts were used to assess the impact of the evaluated polymers on cell viability. Cells were obtained from a healthy patient after signing informed consent, as approved by the local Research Ethics Committee, and registered in the National Research Ethics System (CAAE: 41445320.4.0000.5347). The explants were cultured in Dulbecco Modified Eagle Medium (DMEM) supplemented with 10% fetal bovine serum (FBS) and 1% penicillin, at 37°C and 5% CO_2. Cell viability assays were conducted in 96-well plates. The cells were seeded at a concentration of 5 × 103 and allowed to attach for 24 hours. As an indirect method was used, the extracts were obtained from the immersion of samples (4-mm diameter × 1 mm thick) in 1 mL of DMEM for 24 hours at 37°C. Cells were kept in contact with extracts for 72 hours, and the viable proteins were then stained with 0.4% sulforhodamine B (SRB; n = 5) after fixation with 50% trichloroacetic acid. The stained cells were suspended with 10% Trizma buffer (Sigma), and absorbance was detected at 560 nm in a microplate spectrophotometer (Multiskan GO, Thermo Fisher Scientific). The metabolic activity of treated cells was evaluated with an MTT (3-(4,5-dimethylthiazol-2-yl)-2,5-diphenyltetrazolium bromide) assay (n = 5). A 0.05-mg/mL MTT solution was added to the wells after 69 hours of treatment. Cells were allowed to metabolize the MTT solution for 3 hours before the purple formazan crystals were detected at 590-nm absorbance. Cells grown in DMEM for the same amount of time with no contact with the polymers were analyzed in both assays to normalize the data for the percentage of viable cells.

Statistical Analysis

Data from the evaluated material properties were statistically analyzed using Student *t* test (α = 005). The σ_f and *E* values were also subjected to Weibull analysis.

RESULTS

Table 1 presents σ_f and *E* values for the materials evaluated at baseline (48 hours) and after 6 months of water storage. All comparisons were statistically different ($P = .0000$), except for the mean σ_f values for the AM polymer comparing baseline and 6 months in water ($P = .12$), indicating that the strength did not change based on the storage condition.

Table 2 presents the Weibull modulus (*m*) and characteristic strength (s_0) estimations from σ_f data. The s_0

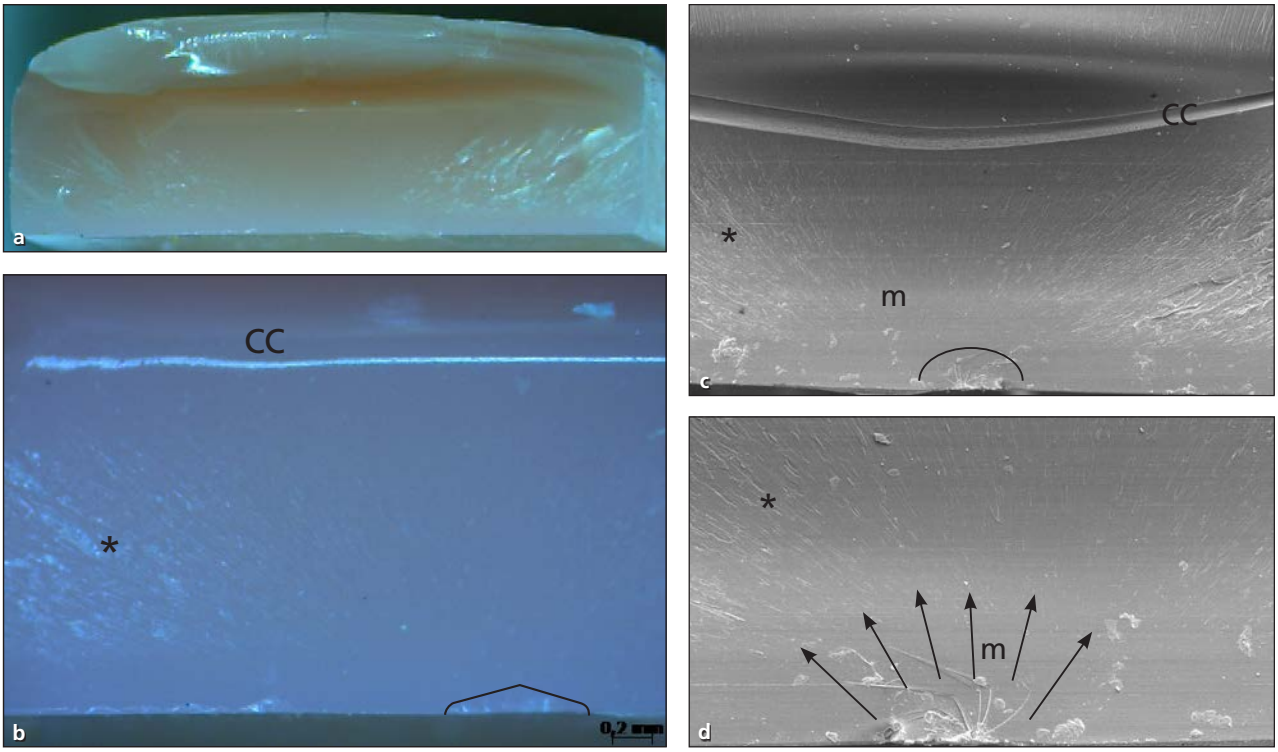

Fig 2 *(a and b)* Stereomicroscopy and *(c and d)* SEM images of an AM specimen tested at baseline. *(a)* Entire fractured surface (×10 magnification). *(b)* Fracture origin *(semicircle)*, hackle region and direction of crack propagation *(*)*, and compression curve (CC) at x32 magnification. *(c)* Fracture origin *(semicircle)*, mirror (m) and hackle *(*)* region, and CC at ×160 magnification. *(d)* Fracture origin and mirror (m) and hackle (*) regions at ×340 magnification. The *black arrows* show the direction of crack propagation. Printing layers *(horizontal lines)* can be observed, inducing the propagation direction of some hackles initiating from the fracture origin.

values for AR decreased significantly after 6 months of water storage, which was not the case for the AM polymer, which was not affected by the 6-month water storage. The material *m* values did not change after 6 months of water storage, maintaining their structural reliability. Yet, the *m* and s_0 values for AR were significantly higher than the values for the AM polymer.

The qualitative evaluation of the fracture surface (fractography), initially performed under stereomicroscopy and later using SEM, showed characteristics (fracture origin, mirror region, direction of fracture propagation, and compression curve) of brittle behavior (Figs 2 to 5).

Regarding Wsp and Wsl (Table 3), AR presented Wsp = 24.9 μg/mm^3, which was higher than the AM polymer (Wsp = 15.2 μg/mm^3) (P = .0000). No solubility was found for AR, while the AM polymer showed a Wsl of 2.3 μg/mm^3 (P = .0000), which is slightly above the standard value.

Biocompatibility was evaluated using SRB and MTT assays (Fig 6). The SRB assay showed a lower percentage of viable cells for AR (69.93% ± 10.55%; 95% CI: 60.68 to 79.17) than for AM (86.27% ± 7.80%; 95% CI: 79.43 to 93.10) (P = 0.023). This finding was confirmed by the MTT assay, which showed higher cell viability for AM (100% ± 12.37%; 95% CI: 88.84 to 111.15) than for AR (80.73% ± 8.25%; 95% CI: 73.49 to 87.96) (P = .010).

DISCUSSION

This study aimed to evaluate the maximum σ_f, E, Wsp, Wsl, and biocompatibility of polymer materials used for complete denture base manufacturing via 3D printing and the conventional method (heat-curing AR). The results showed significant differences between the AM polymer and AR, with the AM polymer showing inadequate σ_f and Wsl values, rejecting the study hypothesis.

The σ_f and E tests were performed based on the ISO standard[24] and previous studies[18,26] for the AM polymer, which is not yet covered by international standards. There are several relevant parameters that should be considered in AM, such as printing orientation, which is still a controversial issue.[26–28] Thus, the present study followed previous studies,[26,28] and the specimens used for σ_f and E were printed at 0 degrees relative to the printing platform, while the compressive load was applied perpendicularly (90 degrees) to the deposition of layers in constructing the specimens.

The influence of printing parameters on σ_f and structural accuracy of AM objects has been evaluated.[26–28]

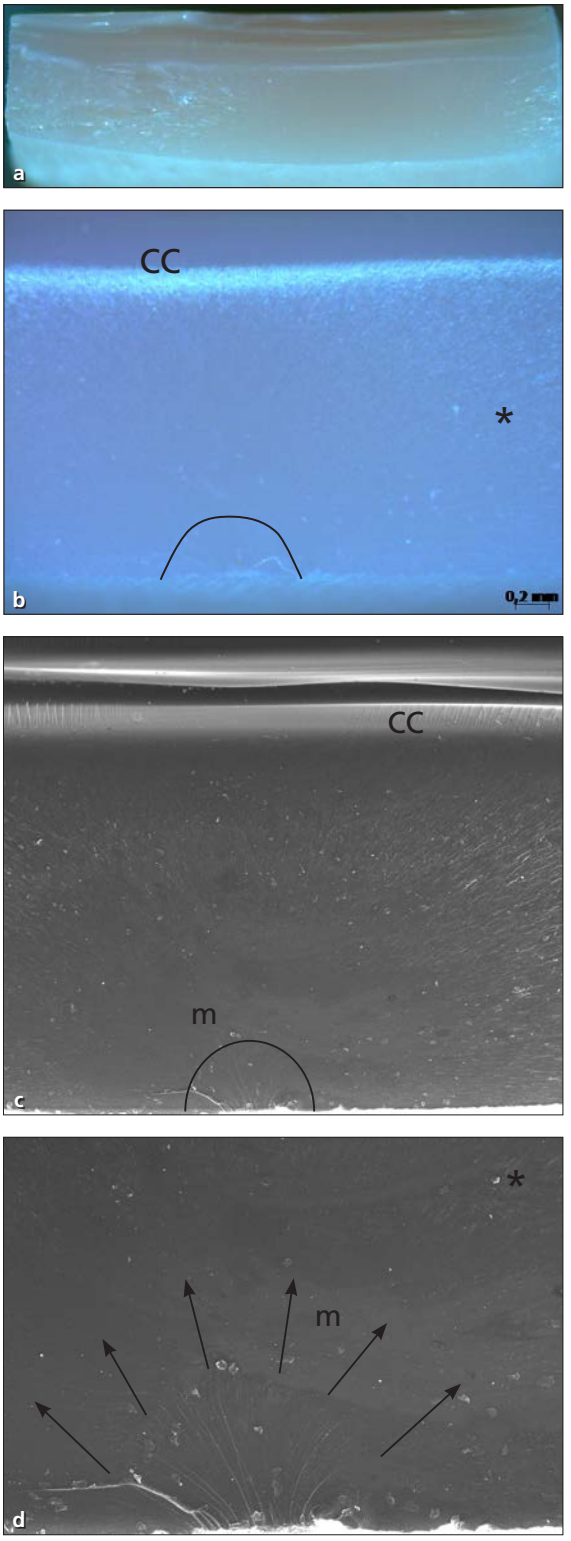

Fig 3 *(a and b)* Stereomicroscopy and *(c and d)* SEM images of an AM specimen tested after 6 months of water storage. *(a)* Entire fractured surface (×10 magnification). *(b)* Fracture origin *(semicircle)*, hackle region and direction of crack propagation *(*)*, and CC at ×32 magnification. *(c)* Fracture origin *(semicircle)*, mirror (m) and hackle (*) regions, and CC at ×160 magnification. *(d)* Fracture origin, direction of crack propagation *(black arrows)*, and mirror (m) and hackle regions at ×340 magnification.

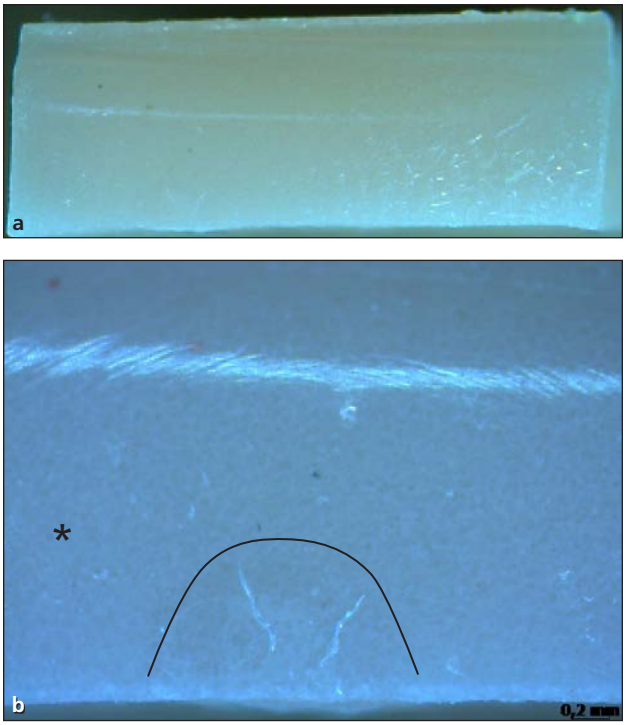

Fig 4 Stereomicroscopic images of an AR specimen at baseline. *(a)* Entire fractured surface (×10 magnification). *(b)* Fracture origin *(semicircle)*, hackle region and direction of crack propagation *(*)*, and CC at ×32 magnification.

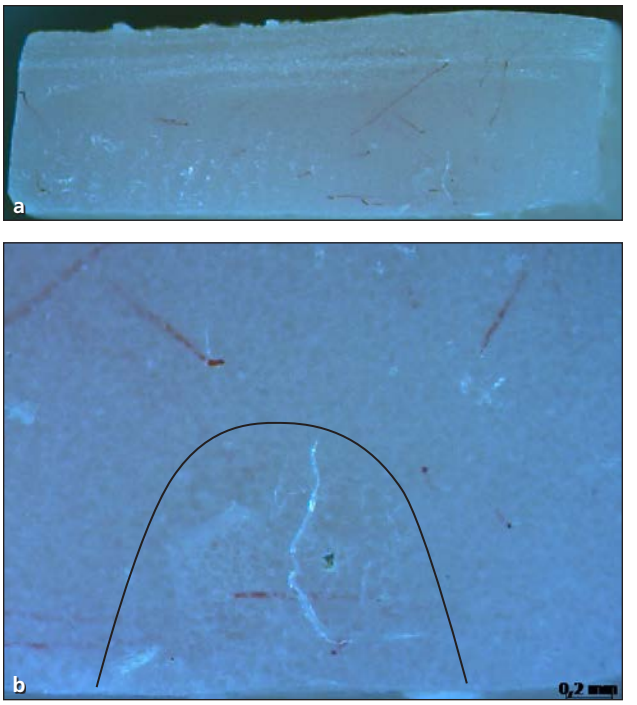

Fig 5 Stereomicroscopic images of an AR specimen after 6 months of water storage. *(a)* Entire fractured surface (×10 magnification). *(b)* Fracture origin *(within the semicircle)* at ×32 magnification.

A study reported that samples printed at 90 degrees (layers parallel to the axial load) showed superior σ_f and *E*, while a 45-degree orientation relative to the printing platform presented the best accuracy.[27] Other studies[26,28] presented divergent results, with specimens constructed at 0 degrees (load applied perpendicularly to the construction of the layers) exhibiting superior mechanical properties compared to those printed at 90 degrees,[26,28] mostly because the adhesion between layers is lower than the intrinsic strength of each layer. This anisotropy is closely associated with the adhesion strength between the printed layers, which can thus change the effect of the printing orientation. Moreover, the adhesive strength can be affected by the type of polymer, the size of the adhesion area, and the polymerization rate, which can also affect the σ_f of the entire structure.[26] This rationale explains the reason for printing the specimens used for σ_f and *E* at 0 degrees and for applying the load perpendicularly to the layer impression, which is similar to the application of biting forces.

The positioning of the structure on the printing platform can also affect its geometry. Objects placed on the edges of the printing platform might be particularly prone to inaccuracies.[27] Therefore, in this study, the specimens were printed in the center of the printing platform.

According to the ISO standard,[24] the σ_f for denture base materials should be at least 65 MPa, which was fully achieved by the AR (78.1 ± 5.5 MPa) but was not reached by the AM (34.4 ± 7.5 MPa) material. However, the literature shows significant variability for the mean σ_f value, [23,26–28] which is significant depending on the printing parameters and postcuring time.[23] In addition, no previous study has tested specimens simulating the oral environment; that is, in distilled water at 37°C.

The σ_f and *E* of two heat-curing ARs used for denture bases, Rodex (S.P.D.) and Lucitone 199 (Dentsply), were evaluated after storage in 37°C artificial saliva for 2 weeks,[18] showing the following results: σ_f = 70 MPa and *E* = 1,600 MPa for Rodex; σ_f = 82 MPa and *E* = 1,230 MPa for Lucitone 199. These results are similar to the AR results of the present study (σ_f = 78 MPa and *E* = 1,700 MPa). However, another study[16] evaluated a conventional acrylic resin (Lucitone 199) under dry conditions and reported mean σ_f values of 116 MPa and *E* values of 2,918 MPa, which are higher than the values from the present study, most likely due to laboratory testing conditions (dry vs wet conditions). The influence of humidity on the σ_f of heat-curing ARs has been reported.[29,30] AR specimens stored in artificial saliva were evaluated after 2, 60, 90, and 120 days and showed a significant reduction in strength (86 to 76 MPa) with storage time (2 to 120 days).[30] Similarly, AR specimens stored in artificial saliva or in water were evaluated for σ_f after 15, 30, 60, and 120 days. A significant reduction in strength (98 to 86 MPa) with increased storage time was reported regardless of the medium (water or artificial saliva).[29] These results were corroborated by another recent study.[23] Such data support the findings of the present study, in which the AR showed a higher mean σ_f value (78 MPa) at baseline than after 6 months of water storage (53 MPa). These studies show the influence of a humid environment on the σ_f of acrylic resins, which is also a frequent finding for dental polymers.[29,30]

Few studies have compared the σ_f of denture base resin structures manufactured by different techniques (conventional vs AM). Conventionally manufactured AR specimens showed mean σ_f values from 62 to 110 MPa,[22] in

Table 3 Water Sorption (Wsp) and Water Solubility (Wsl) for Each Material

		Material	
Property	n	AR	3D
Wsp (μg/mm³)	5	24.91 ± 0.55 (24.2–25.6)^{A}	15.16 ± 0.57 (14.4–15.8)^{B}
Wsl (μg/mm³)	5	0.00 ± 0.00^{A}	2.33 ± 0.48 (1.7–2.9)^{B}

Data are reported as mean ± SD (95% CI).
Different uppercase letters indicate statistical differences in the same line ($P \leq .05$).

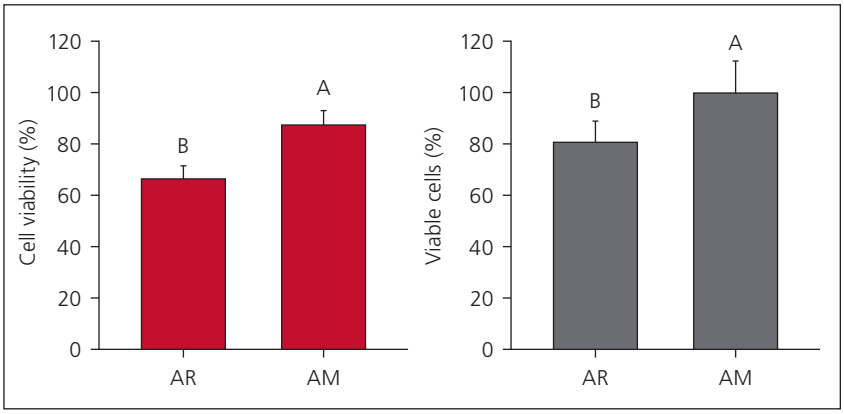

Fig 6 Mean and SD values of cell viability obtained using SRB and MTT assays. Different capital letters indicate a statistically significant difference between materials according to Student *t* test.

agreement with the present results, while the AM polymer (NextDent Base) showed mean σ_f values from 60 to 84 MPa,[22] which are higher than the values obtained in the present study (σ_f = 34 MPa). This difference can be explained by the different testing conditions (dry vs wet). Furthermore, it has been reported that the longer the postcuring time of the AM structure, the higher the σ_f.[23] The mean σ_f values of an AM resin (Denture Base, Dentca) ranged from 50 to 127 MPa for 5 and 20 minutes of postcuring, respectively. Similar variation was observed for the *E* values.[23]

It is worth noting that the present study simulated oral conditions to evaluate mechanical properties (σ_f and *E*), and no study was found in the literature reporting the influence of water storage at 37°C for 6 months with subsequent testing of these properties in 37°C water. These experimental conditions can approximate the laboratory data to the expected clinical results.

Visual and low-magnification microscopy (optical microscopy and stereomicroscopy) often reveal basic characteristics of fractured surfaces.[31] Yet, some materials and structures require higher magnifications (SEM) to define and measure relevant fractographic features, such as the fracture origin. Nevertheless, quantitative fracture strength data should be supported by a detailed investigation of the fractured surface using fractographic principles.[32] Such rationale was applied in the present study; the fractured surfaces were examined using stereomicroscopy and SEM (see Figs 2 to 5). Fractography showed that the mirror region (m) was usually smaller for specimens tested at baseline, indicating higher fracture stress and, consequently, higher mean σ_f and *E* values than the specimens aged for 6 months in water (see Table 1). Water aging is deleterious for most materials, including polymers,[29,30] and impairs the mechanical properties (σ_f and *E*). Hence, aged structures accumulate less energy up to the catastrophic fracture. These observations are consistent with the quantitative (flexure test) and qualitative (fractography) analyses of the structures in the present study. Further, no study has yet evaluated the fractured surface of denture base structures based on fractographic principles, although two studies evaluated the impact resistance using the Charpy method, followed by a morphologic evaluation of denture base AR. As a consequence of the impact test, these studies had difficulties showing the fractured surface characteristics usually observed in bending tests.[33,34] Another study used SEM and profilometry to assess the topography of the fractured surface.[23]

The solubility of ARs represents the amount of water-soluble ingredients (eg, monomers that do not participate in the reaction, such as plasticizers and initiators) that migrate to water during the immersion time of the experiment. As residual monomers may get into oral fluids and irritate the oral mucosa,[20] these materials must have low or no solubility. According to the ISO standard,[24] the maximum Wsl value for heat-curing (Type 1) and light-activated polymers (Type 4, by analogy to AM polymers) for complete denture bases is 1.6 μg/mm^3, and the maximum Wsp value for these materials is 32 μg/mm^3. In the present study (see Table 2), AR showed a higher Wsp (24.9 μg/mm^3) than the AM polymer (15.2 μg/mm^3), but both values are below the maximum Wsp.[24] AR showed no solubility, and the AM polymer showed a Wsl of 2.3 μg/mm^3. Yet, previous studies on commercially available ARs showed values of Wsp = 20.3 μg/mm^3 and Wsl = 3.02 μg/mm^3 for Lucitone 199[19] and Wsp = 38.3 μg/mm^3 and Wsl = 0.18 μg/mm^3 for Lucitone 550.[20] Only one study[23] evaluated the Wsp and Wsl of AM denture base resins, with Wsp ranging between 23 and 26 μg/mm^3 and Wsl ranging from 2 to 6 μg/mm^3, which are slightly greater than the values found in the present study.

There are different methods to assess the biocompatibility or the cytotoxicity of biomaterials. MTT is often used to investigate cell metabolism in vitro.[35–39] Additionally, the impact on cell structure is often assessed by SRB analysis.[39,40] Despite the solubility observed for the AM material, its cell viability was higher than AR in both MTT and SRB analyses. While a reduction in cell structure was observed in both groups, the maintenance of cell metabolism was found for the AM material in MTT. Further, a study[23] reported that longer postcuring time increased the cell viability of the AM resin.

Although SLA and digital light processing (DLP) are the most popular AM technologies in dentistry,[21] it is worth noting that they have differences. SLA is done in incremental curing that traces the object, while in DLP and LCD technology, the curing process happens in a full surface cure. Using one AM technology (SLA) and one printing resin can be considered limitations of the present study. Future studies should investigate different AM technologies and printing resins, and the influence of additional parameters, such as postcuring, on the long-term behavior of AM denture base materials.

CONCLUSIONS

Significant differences were found for the evaluated material properties of the polymers used for denture base fabrication. The σ_f, *E*, and Wsp were higher for the AR than for the AM polymer, and the Wsl was higher for the AM polymer than for the AR, rejecting the study hypothesis. Water storage for 6 months had a negative influence on σ_f and *E* for AR. It can be concluded that the AM polymer is biocompatible but requires further development to improve other material properties, such as σ_f and Wsl.

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REFERENCES

1. Tyrovolas S, Koyanagi A, Panagiotakos DB, et al. Population prevalence of edentulism and its association with depression and self-rated health. Sci Rep 2016;6:37083.
2. Sheiham A, Steele JG, Marcenes W, Tsakos G, Finch S, Walls AW. Prevalence of impacts of dental and oral disorders and their effects on eating among older people; A national survey in great britain. Community Dent Oral Epidemiol 2001;29:195–203.
3. Musacchio E, Perissinotto E, Binotto P, et al. Tooth loss in the elderly and its association with nutritional status, socio-economic and lifestyle factors. Acta Odontol Scand 2007;65:78–86.
4. Papadaki E, Anastassiadou V. Elderly complete denture wearers: A social approach to tooth loss. Gerodontology 2012;29:e721–e727.
5. Fiske J, Davis DM, Frances C, Gelbier S. The emotional effects of tooth loss in edentulous people. Br Dent J 1998;184:90–93.
6. Kattadiyil MT, AlHelal A, Goodacre BJ. Clinical complications and quality assessments with computer-engineered complete dentures: A systematic review. J Prosthet Dent 2017;117:721–728.
7. Van noort R. The future of dental devices is digital. Dent Mater 2012;28:3–12.
8. Chung YJ, Park JM, Kim TH, Ahn JS, Cha HS, Lee JH. 3D printing of resin material for denture artificial teeth: Chipping and indirect tensile fracture resistance. Materials (Basel) 2018;11:1798.
9. Cha HS, Park JM, Kim TH, Lee JH. Wear resistance of 3D-printed denture tooth resin opposing zirconia and metal antagonists. J Prosthet Dent 2020;124:387–394.
10. Park C, Kee W, Lim HP, Park SW. Combining 3D-printed metal and resin for digitally fabricated dentures: A dental technique. J Prosthet Dent 2020;123:389–392.
11. Kessler A, Hickel R, Reymus M. 3D printing in dentistry-state of the art. Oper Dent 2020;45:30–40.
12. Diaz-Arnold AM, Vargas MA, Shaull KL, Laffoon JE, Qian F. Flexural and fatigue strengths of denture base resin. J Prosthet Dent 2008;100:47–51.
13. Pfeiffer P, Rosenbauer EU. Residual methyl methacrylate monomer, water sorption, and water solubility of hypoallergenic denture base materials. J Prosthet Dent 2004;92:72–78.
14. Anusavice KJ, Shen CS, Rawls HR. Prosthetic polymers and resins. In: Science of Dental Materials, ed 12. Elsevier, 2013:474–498.
15. Basso GR, Moraes RR, Borba M, Griggs JA, Della Bona A. Flexural strength and reliability of monolithic and trilayer ceramic structures obtained by the CAD-on technique. Dent Mater 2015;31:1453–1459.
16. Aguirre BC, Chen JH, Kontogiorgos ED, Murchison DF, Nagy WW. Flexural strength of denture base acrylic resins processed by conventional and CAD-CAM methods. J Prosthet Dent 2020;123:641–646.
17. Prpić V, Slacanin I, Schauperl Z, Ćatić A, Dulčić N, Čimić S. A study of the flexural strength and surface hardness of different materials and technologies for occlusal device fabrication. J Prosthet Dent 2019;121:955–959.
18. Ajaj-Alkordy NM, Alsaadi MH. Elastic modulus and flexural strength comparisons of high-impact and traditional denture base acrylic resins. Saudi Dent J 2014;26:15–18.
19. Dhir G, Berzins DW, Dhuru VB, Periathamby AR, Dentino A. Physical properties of denture base resins potentially resistant to candida adhesion: Basic science research. J Prosthodont 2007;16:465–472.
20. Cucci AL, Vergani CE, Giampaolo ET, Afonso MC. Water sorption, solubility, and bond strength of two autopolymerizing acrylic resins and one heat-polymerizing acrylic resin. J Prosthet Den 1998;80:434–438.
21. Della Bona A, Cantelli V, Britto VT, Collares KF, Stansbury JW. 3D printing restorative materials using a stereolithographic technique: A systematic review. Dent Mater 2021;37:336–350.
22. Prpić V, Schauperl Z, Ćatić A, Dulčić N, Čimić S. Comparison of mechanical properties of 3D-printed, CAD/CAM, and conventional denture base materials. J Prosthodont 2020;29:524–528.
23. Aati S, Akram Z, Shrestha B, et al. Effect of post-curing light exposure time on the physico-mechanical properties and cytotoxicity of 3D-printed denture base material. Dent Mater 2022;38:57–67.
24. International Organization for Standardization. ISO 20795-1:2013. Dentistry—Base polymers. ISO, 2013.
25. International Organization for Standardization. ISO 3696:1987. Water for analytical laboratory use—Specification and test methods. ISO, 1987.
26. Shim JS, Kim JE, Jeong SH, Choi YJ, Ryu JJ. Printing accuracy, mechanical properties, surface characteristics, and microbial adhesion of 3D-printed resins with various printing orientations. J Prosthet Dent 2020;124:468–475.
27. Unkovskiy A, Bui PHB, Schille C, Geis-Gerstorfer J, Huettig F, Spintzyk S. Objects build orientation, positioning, and curing influence dimensional accuracy and flexural properties of stereolithographically printed resin. Dent Mater 2018;34:324–333.
28. Alharbi N, Osman R, Wismeijer D. Effects of build direction on the mechanical properties of 3D-printed complete coverage interim dental restorations. J Prosthet Dent 2016;115:760–767.
29. Jagini AS, Marri T, Jayyarapu D, Kumari R, Vidhyadhar D, Manisha K. Effect of long-term immersion in water and artificial saliva on the flexural strength of two heat cure denture base resins. J Contemp Dent Pract 2019;20:341–346.
30. Chandrahari N, Ravi Kumar C, Salgar AR, Prathibha N, Singh M, Singh S. Comparison of fracture resistance of heat cure resins polymerized by conventional and microwave methods after immersion in artificial saliva. J Contemp Dent Pract 2019;20:71–77.
31. Quinn G. Recommended practice guide: Fractography of ceramics and glasses. 3rd ed, special publication (NIST SP), National Institute of Standards and Technology, Gaithersburg, MD [online] 2020.
32. Della Bona A. Bonding to ceramics: Scientific evidences for clinical dentistry. São Paulo: Artes Medicas, 2009.
33. Faot F, Costa MA, Del Bel Cury AA, Rodrigues Garcia RC. Impact strength and fracture morphology of denture acrylic resins. J Prosthet Dent 2006;96:367–373.
34. Praveen B, Babaji HV, Prasanna BG, Rajalbandi SK, Shreeharsha TV, Prashant GM. Comparison of impact strength and fracture morphology of different heat cure denture acrylic resins: An in vitro study. J Int Oral Health 2014;6:12–16.
35. Goiato MC, Freitas E, dos Santos D, de Medeiros R, Sonego M. Acrylic resin cytotoxicity for denture base–Literature review. Adv Clin Exp Med 2015;24:679–686.
36. Ata SO, Yavuzyilmaz H. In vitro comparison of the cytotoxicity of acetal resin, heat-polymerized resin, and auto-polymerized resin as denture base materials. J Biomed Mater Res B Appl Biomater 2009;91:905–909.
37. Çakırbay Tanış M, Akay C, Sevim H. Cytotoxicity of long-term denture base materials. Int J Artif Organs 2018;41:677–683.
38. Raszewski Z. Influence of polymerization method on the cytotoxicity of three different denture base acrylic resins polymerized in different methods. Saudi J Biol Sci 2020;27:2612–2616.
39. Britto VT, Cantelli V, Collares FM, Bertol CD, Della Bona A. Biomechanical properties of a 3D printing polymer for provisional restorations and artificial teeth. Dent Mater 2022;38:1956–1962.
40. van Tonder A, Joubert AM, Cromarty AD. Limitations of the 3-(4,5-dimethylthiazol-2-yl)-2,5-diphenyl-2H-tetrazolium bromide (MTT) assay when compared to three commonly used cell enumeration assays. BMC Res Notes 2015;8:47.