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The Effects of Different Chemical Disinfectants on the Strength, Surface, and Color Properties of Conventional and 3D-Printed Fabricated Denture Base Materials

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Abstract: Objectives: The disinfection of fabricated prostheses is crucial to prevent cross-infection between dental laboratories and clinics. However, there is a lack of information about the effects of chemical disinfection on 3D-printed denture base resins. This study aimed to evaluate the impact of different disinfectants on the flexural strength, elastic modulus, micro-hardness, surface roughness (Ra), and change in color of 3D-printed and conventional heat-polymerized (HP) denture base resins (DBRs). Methods: A total of 240 specimens (80 bar-shaped ($64 \times 10 \times 3.3$ mm) and 160 disk-shaped (10×2 mm)) were made from HP and 3D-printed DBRs. For each resin, the specimens were divided into four groups ($n = 10$) according to the disinfectant solution. One remained in water without disinfection as a control group, while the other three groups were disinfected using 1% sodium hypochlorite, 2% glutaraldehyde, or 10% Micro 10+ for 30 min. The flexural strength, elastic modulus, micro-hardness, Ra, and color change were measured. The collected data were statistically analyzed using a two-way ANOVA and Tukey's post hoc test ($\alpha = 0.05$). Results: A significant decrease in flexural strength, elastic modulus, and hardness was found with sodium hypochlorite ($p < 0.05$). When comparing the resins per solution, the 3D-printed resin showed a significant decrease in flexural strength, elastic modulus, and hardness compared with PMMA ($p < 0.001$), while no change was found in the Ra of both resins with all disinfectants ($p > 0.05$). Disinfecting with sodium hypochlorite resulted in a significant increase in color change for both resins ($p < 0.05$); however, all the changes were within clinically acceptable limits. Sodium hypochlorite showed the highest color change, while 2% glutaraldehyde and 10% Micro 10+ showed no significant changes in the tested properties ($p > 0.05$). Conclusions: Neither resin showed a change in surface roughness with immersion in disinfectants. Sodium hypochlorite had an adverse effect on the flexural properties, hardness, and change in color of the PMMA and 3D-printed DBRs, while the other disinfectants had no effect on the tested properties.

Keywords: 3D printing; disinfectants; mechanical testing; surface properties; denture base; acrylic resin

1. Introduction

Edentulism, or complete tooth loss, is a very challenging condition that has a negative impact on quality of life [1]. Complete denture (CD) fabrication is the conventional rehabilitation of edentulism and is still a feasible treatment strategy [1,2]. The material of choice for CD fabrication is polymethylmethacrylate (PMMA), and this is attributed to the following reasons: first, it has a lower cost compared to other options; second, it has superior physical and mechanical properties, with an excellent esthetic appearance; and finally, it affords better manipulation and handling for any dental technician [3]. On the other hand, this material reveals some dimensional and color changes, with a possibility of fracture in spite of its mechanical and physical properties; moreover, a CD, if not properly cleaned and maintained by the patient, can cause Candidal infection or even tissue abrasions and irritation [4]. One major factor that could really affect the mechanical and esthetic properties of this material is surface and subsurface voids [5]. Thus, paying attention to and obtaining superior surface characteristics during denture fabrication can pave the way for better esthetics and higher longevity of the denture base [2].

Digital technology has become increasingly popular in several dental specialties in recent years. Nevertheless, the use of digital tools, materials, and computer-aided design and computer-assisted manufacturing (CAD/CAM) in the design and production of dental prostheses has helped to lessen the workload for dental technicians and dentists [6]. Two methods have been designed for denture fabrication: the subtractive method, in which the denture is milled from a prefabricated resin disc, and the additive method, in which the denture is built up by three-dimensional printing (3D printing) using fluid resins [3]. The benefits of 3D-printed dentures include shorter production times, more accuracy, lower costs, fewer patient visits, and greater patient comfort [7]. The primary disadvantages of subtraction procedures are the high cost and waste of milling machines, burs, and restorative materials, as well as the device's restricted motion range to make complicated forms [8].

Different types of technologies for denture fabrication have been reported, such as stereolithography (SLA), digital light projection (DLP), and photopolymer jetting (PolyJet) [9]. Among these methods, DLP is considered to be superior when compared to the others because of its lower material consumption, faster speed, and greater precision [10]. In term of the fabrication method effect, 3D-printed DBRs show comparable properties to conventional PMMA [3], while 3D-printed DBRs show superior properties compared to PMMA regarding accuracy, fit, and adaptability; therefore, 3D printing is recommended for DBR fabrication [5]. According to a recent review report, printed dentures have problems with strength, color stability, and stainability; however, these issues could be resolved by using new materials and modifying existing technology [11]. With growing evidence supporting the benefits of 3D printing technologies, several studies have been conducted to assess the performance of dentures printed with different printing technologies. In terms of strength, 3D-printed dentures show low strength but are still within the ISO recommendations [12]; in addition, they also have poor surface properties [4,11,13]. To obtain the benefits from additive technologies with optimum properties, controlling the printing factors (pre-printing, printing, and/or post-printing parameters) is suggested [14]. By modifications to these parameters, improvements in strength, surface properties, and antimicrobial efficacy were achieved [8,14,15].

Prosthetics produced in dental laboratories are vulnerable to microbial contamination during the manufacturing process [16]. Consequently, various contamination sources are reported, including contact with contaminated hands, felt disks and pumice used in the polishing process, and machines and equipment used for denture base resin (DBR) fabrication. Additionally, the fabrication process involves several steps, most of which are conducted in a laboratory environment with the risk of prosthesis contamination [17,18]. Other sources of contamination include when the prosthesis is returned from the dental office for repair, relining, or rebasing after patient use [19]. According to the literature, significant microbiological cross-contamination can occur when transferring prostheses between dental offices and dental laboratories [18]. Therefore, less attention has been paid to disinfecting dentures; instead, the focus should be on preventing cross-contamination through infection-control procedures, including the barrier technique, sterilization, and disinfection of the dental office and its equipment. A dental prosthesis provides a conduit for organisms to be transferred from patients to laboratory and dental staff [19,20]. Dentures must be disinfected to prevent cross-contamination and enhance cleanliness. If proper disinfection measures are not followed, the dental office–prosthesis laboratory connection could be a cross-infection conduit [18].

Several disinfectants have been suggested at different concentrations and durations with variations in disinfection level [16]. However, suitable disinfection should be effective without deteriorating the prosthesis structures and properties [21]. Among the common disinfectants, sodium-hypochlorite- and glutaraldehyde-based disinfectants are often used in dentistry, and aldehyde-free disinfectants are commonly used for complete microbial elimination from disinfected prostheses [16,22–24].

Because the restorations are exposed to temperature changes and functional stress during their clinical service, reversible elastic deformation, irreversible plastic deformation, and fracture [9] are all possible outcomes of residual stresses created by these dynamic changes [25]. Furthermore, a full denture's color stability may be reduced by its surface roughness [8]. CD discoloration may be a sign of aging and material degradation [25], which may ultimately necessitate denture replacement [7].

Before considering 3D-printed dentures as a good substitute for traditional PMMA dentures, their mechanical, physical, and cosmetic qualities should be carefully examined. Therefore, the present in vitro study aimed to evaluate and compare the flexural strength, modulus of elasticity, micro-hardness, surface roughness, and change in color of heat-polymerized acrylic resin with that of 3D-printed resin after disinfections with various chemical disinfectants. The null hypothesis was that the disinfectants would not affect the tested properties of the tested DBRs.

2. Materials and Methods

2.1. Specimen Preparation

The sample size calculation revealed that $n = 10$ was sufficient to detect effect sizes for the main effects and pairwise comparisons, with the power level set at 80% and 95% confidence [26]. According to each test specification, specimens were fabricated with different dimensions; for flexural properties, rectangular specimens ($64 \times 10 \times 3.3 \pm 0.2$ mm), while surface and color changes were tested using disk-shaped (10×2 mm) specimens.

The test specimens were designed using a software program (SolidWorks version 2024, Dassault Systèmes SolidWorks Corp., Aix-en-Provence, France) and saved as a standard tessellation language (STL). The STL was used to mill the specimens from milling wax blocks (Duo Cad; FSM Dental, Ankara, Türkiye), which were placed inside a metal flask or (split stainless-steel mold with metal slots of required dimensions) produced to obtain PMMA (Temdent Classic; Schütz Dental GmbH, Rosbach, Germany) specimens [27]. As instructed by the manufacturer, the PMMA was hand-blended and packed into the mold followed by polymerization as a conventional method.

For 3D-printed resin, an STL file was used to print 3D-printed specimens. The specimens were printed vertically with 100 µm thick layers (z-direction angulated 90° to the printing direction) with a digital light processing printer (D30 II, Rapid Shape, Heimsheim, Germany) using a fluid resin (FREEPRINT denture, Detax, Ettlingen, Germany) [28]. Following printing, the specimens were post-cured from all sides for 20 min using ultraviolet light (385 nm) with UV-A type 3 in a light box (type E0202; Yizhet, Shenzhen, China) and cleaned with 99% isopropyl alcohol for 5 min as per the manufacturer's instructions. Silicon carbide paper grit P1200 (Paper SiC P1200; Struers GmbH, Rosbach, Germany) was then used to grind all the manufactured specimens to their final dimensions. A digital caliper (Digimatic Micrometer, Mitutoyo, Kanagawa, Japan) was used to validate the specimen dimensions to the closest ±0.02 mm after they had been ground. The approved specimens were then kept for 48 h in distilled water at 37 °C [29].

2.2. Specimen Disinfection Procedures

The prepared specimens (for each main group, conventional and 3D-printed resins) were randomly divided into four groups (n = 10). One was immersed in water as a control, while the other three groups were disinfected using 1% sodium hypochlorite, 2% glutaraldehyde, and 10% Micro 10+ for a specified time based on immersion protocols (Table 1).

Table 1. Immersion solutions.

Solution	Composition	Immersion Protocol
Glutaraldehyde	An organic compound with the formula $(CH_2)_3(CHO)_2$. The molecule consists of a five-carbon chain doubly terminated with formyl groups.	20 min immersion in at least a 2% solution of glutaraldehyde at room temperature.
Sodium Hypochlorite	0.5% Sodium hypochlorite solution, 1% active chlorine.	Solution of 5.25% sodium hypochlorite (1:5 dilution) diluted to obtain 1% sodium hypochlorite by adding 50 mL of sodium hypochlorite to 200 mL of water with iImmersion for 10 min at room temperature.
Micro 10+	Micro 10+ is an aldehyde-free concentrated solution. A total of 100 g of Micro 10+ contains 9 g of alkylbenzyldimethylammonium chloride, amphoteric and non-ionic surfactants, complexant, corrosion inhibitor, and additives.	Very economical 2% dilution. Contact time: 15 min.

2.3. Specimen Testing

The flexure strength and elastic modulus were assessed using a three-point bending test. A load cell with a 5 kN capacity and a 1 mm/min cross-head speed was used to apply the load until the specimen fractured using a universal testing machine (Instron Industrial Products, Model 3345). Stress–strain curves were generated using software (Instron® Bluehill Lite Software, Norwood, MA, USA). The limiting stress at which failure occurs is represented by the flexural strength (FS), calculated using Equation (1):

$$FS\ (\acute{6}) = 3F\ (L)/2wh^2 \quad (1)$$

Here, F is the maximum load at the point of fracture, L is the span, w is the width of the sample, and h is its height. The modulus of elasticity was calculated mathematically from the stress–strain curve obtained during the flexural strength test. The modulus of elasticity (MPa) = stress/strain within the elastic portion [30].

Surface roughness (Ra) measurements were measured on the disk specimens, and a USB digital surface profile gauge (Elcometer 224/2, Elcometer Instruments, Manchester, England) was used to assess surface roughness. The roughness (Ra, µm) was generated as the arithmetic mean between the peaks and valleys recorded after the profilometer needle had scanned a span of 2 mm in length, with a cut-off of 0.25 mm, to optimize filtering and surface undulation. Each surface was read five times, starting from three different positions and always with the needle scanning the specimen's geometric center. The mean roughness of each specimen was determined as the average of the five values.

The micro-hardness (VHN) was assessed using a Vickers hardness tester (ZHU 2.5, Zwick/Roell GmbH, Ulm, Germany). The load was applied using a Vickers indenter with a speed of 1 mm/min at the contact point and a dwell time of 2 s at the loading point. For each specimen, three readings at different points on the specimen surfaces were conducted, followed by an average calculation per specimen.

For color changes (ΔE_{ab}), a reflective spectrophotometer (X-Rite, model RM200QC) was used to measure the colors of all the disk specimens after fabrication and after treatment. The aperture size was set at 4 mm, and the specimens were precisely positioned in relation to the device. Measurements were made against the CIE standard illuminant D65 against a white background using the CIE L*a*b* color space. Equation (2) was used to assess the specimen color changes:

$$\Delta E_{ab} = (\Delta L^{*2} + \Delta a^{*2} + \Delta b^{*2})\ ½ \quad (2)$$

Here, L* = lightness (0–100), a* = the red/green axis, and b* = the yellow/blue axis [31]. The National Bureau of Standards (NBS) was used as a reference for the color change comparison and calculated using the following equation: NBS = $\Delta E_{ab} \times 0.92$. A material is deemed esthetically and clinically acceptable when NBS units fall within the range of 3.7 NBS units. Differences exceeding 3.7 NBS units are classified as a "mismatch" and viewed as clinically unacceptable [31–33].

The data was statistically analyzed using SPSS v28 (IBM, Armonk, NY, USA). The Shapiro-Wilk test and histograms were used to evaluate the normal distribution of data. Data were presented as mean and standard deviation (SD) and were analyzed using a two-way ANOVA test (for the combined effect of disinfectant and material type) followed by a post hoc analysis (Tukey's) test. The significance level was set at $p < 0.05$.

3. Results

Table 2 summarizes the mean values, SD, and significance for the groups regarding flexural strength, elastic modulus, hardness, and surface roughness. For PMMA, a significant difference was found when comparing the disinfectants ($p = 0.04$). For the pairwise comparison, a significant decrease in flexural strength with sodium hypochlorite ($p < 0.05$) was found compared with the other solutions, while no significant difference was found between water, glutaraldehyde, and Micro 10+ ($p > 0.05$). For the 3D-printed DBRs, no significant differences were found between the tested groups ($p = 0.640$). When comparing resins per solution, the 3D-printed resin showed a significant decrease in flexural strength ($p < 0.001$), and the highest flexural strengths were recorded with water immersion (101.3 $\pm$ 4.00 and 89.3 $\pm$ 3.22 MPa for PMMA and 3D-printed resin, respectively).

For both resins, immersion in disinfectant did not produce a significant difference in the elastic modulus compared to water immersion ($p > 0.05$), except with sodium hypochlorite, which showed a significant decrease in the elastic modulus ($p < 0.05$). When comparing resin based on the immersion solution, the 3D-printed resin showed a significantly decreased elastic modulus compared with PMMA, and the highest elastic moduli were reported for water immersion (2954.3 ± 93.77 and 2208.5 ± 87.27 MPa for PMMA and 3D-printed DBRs, respectively).

Table 2. Mean value and significance between groups of both resins with different disinfectants, showing effects for flexural strength, elastic modulus, hardness, and surface roughness properties.

Tested Properties	Immersion Solution	Resins Mean ± SD		
		Heat-Polymerized	3D-Printed	*p* Value
Flexural strength (MPa)	Water	101.3 ± 4.00 [a]	89.3 ± 3.22 [a]	0.000 *
	Glutaraldehyde	99.4 ± 2.244 [a]	86.7 ± 3.76 [a]	0.000 *
	Sodium Hypochlorite	91.5 ± 2.87	82.7 ± 1.61	0.000 *
	Micro 10+	98.1 ± 2.91 [a]	86.9 ± 3.17 [a]	0.000 *
	p value	0.020 *	0.04 *	
Elastic modulus (MPa)	Water	2954.3 ± 93.77 [a]	2208.5 ± 87.27 [a]	0.000 *
	Glutaraldehyde	2920.8 ± 94.4 [a]	2182.5 ± 92.94 [a]	0.000 *
	Sodium Hypochlorite	2514.4 ± 98.71	1975.5 ± 69.12	0.000 *
	Micro 10+	2854.6 ± 123.99 [a]	2105.5 ± 143.06 [a]	0.000 *
	p value	0.032 *	0.010 *	
Hardness (VHN)	Water	17.38 ± 0.35 [a]	16.8 ± 0.44 [a]	0.73
	Glutaraldehyde	17.25 ± 0.33 [a]	16.58 ± 0.47 [a]	0.08
	Sodium Hypochlorite	14.02 ± 0.39	13.27 ± 0.31	0.10
	Micro 10+	17.11 ± 0.25 [a]	16.52 ± 0.41 [a]	0.53
	p value	0.03 *	0.04 *	
Roughness (Ra, µm)	Water	0.155 ± 0.035	0.399 ± 0.029	0.000 *
	Glutaraldehyde	0.159 ± 0.026	0.405 ± 0.035	0.000 *
	Sodium Hypochlorite	0.167 ± 0.014	0.414 ± 0.032	0.000 *
	Micro 10+	0.160 ± 0.015	0.411 ± 0.032	0.000 *
	p value	0.174	0.091	

* Significant at p value < 0.05. Same small letter vertically per column indicates insignificant difference pairwise comparison.

For both resins, immersion in disinfectant showed no significant difference in hardness compared to water immersion ($p > 0.05$), except with sodium hypochlorite, which showed a significant decrease in hardness ($p < 0.05$), leading to the lowest hardnesses (14.02 ± 0.39 and 13.27 ± 0.31 VHN for PMMA and 3D-printed resin, respectively). When comparing the resins based on the immersion solution, the 3D-printed resin showed insignificantly low hardnesses compared with PMMA ($p > 0.05$).

The immersion of the PMMA and 3D-printed DBRs in disinfectant showed no significant difference in surface roughness ($p = 0.174$ and $p = 0.091$ for PMMA and 3D-printed DBRs, respectively). When comparing the resins based on the immersion solution, the 3D-printed resin showed a significantly increased surface roughness ($p < 0.001$).

Table 3 shows the mean values, SD, and significance of the color changes. Immersion of the specimens in sodium hypochlorite significantly increased the color change compared with glutaraldehyde and Micro 10+ ($p < 0.05$), with no significant difference in color change between glutaraldehyde and Micro 10+ ($p > 0.05$). When comparing the resins based on the immersion solution, no significant differences were found in the color change between the PMMA and 3D-printed DBRs ($p > 0.05$). The highest NBS value (1.04) was recorded with sodium hypochlorite, which was lower than 3.7, revealing that all changes were within the clinically acceptable value.

Table 3. Change in color (ΔE_{ab}) of tested resins after disinfectants.

Immersion Solutions	Resin and NBS Unit Mean ± SD				
	Heat-Polymerized	**NBS**	**3D-Printed**	**NBS**	***p* Value**
Glutaraldehyde	0.61 ± 0.09 [a]	0.56	0.68 ± 0.10 [a]	0.62	0.30
Sodium Hypochlorite	1.13 ± 0.09	1.04	1.67 ± 0.06	1.53	0.09
Micro 10+	0.71 ± 0.06 [a]	0.65	0.78 ± 0.09 [a]	0.71	0.11
p value	0.001 *		0.001 *		

* $p < 0.05$ set as significant level. Same small letter vertically per column indicates insignificant difference pairwise comparison. The National Bureau of Standards (NBS) values deemed esthetically and clinically acceptable when falling within the range of 3.7, while values greater than 3.7 are classified as a mismatch (clinically unacceptable).

4. Discussion

This study was conducted to investigate the effect of chemical disinfectants on the flexural strength, elastic modulus, hardness, surface roughness, and change in color of 3D-printed DBRs compared with PMMA DBRs. The null hypothesis was partially rejected, as all the disinfectant solutions significantly impacted all the tested properties except roughness, which showed no significant change.

Denture disinfection is a mandatory process to avoid cross-contamination between dental offices and dental laboratories where the prosthesis is fabricated, adjusted, repaired, and rebased [16,34,35]. With advanced technology for removable prosthesis fabrication, no study has investigated the effect of the disinfection process on the properties of 3D-printed DBRs. Thus, one 3D-printed resin was selected for investigation in the present study compared with one conventional PMMA DBR. Two commonly used disinfectant solutions were selected, sodium hypochlorite and glutaraldehyde, as well as one aldehyde-free disinfectant (Micro 10+), and the three selected disinfectants have strong antimicrobial activities [36]. An infection-control procedure designed to avoid cross-contamination was assessed in preliminary studies [36,37]. The findings showed that 4% chlorhexidine gluconate solutions and sodium hypochlorite solutions reduced the microbial growth on the dentures in 10 min [36,37].

The literature reports significant fluctuation in the use of several disinfectants regarding concentration and duration [16,20]. The most appropriate disinfectant should meet most of the ideal agent's criteria while retaining the prosthesis structure [21]. Sodium hypochlorite is a commonly used disinfectant and has a wide range of activity within a short disinfectant period [17,18]. Rodrigues et al. and Salvia et al. reported immersion in sodium hypochlorite containing 2% active chloride for 30 min as the most efficient approach for disinfecting acrylic resin prostheses [38,39]. Furthermore, Chau et al. [23] reported strong disinfectant activities of 1% sodium hypochlorite in removing microbes from denture surfaces. Despite its effectiveness as a disinfectant, sodium hypochlorite has numerous disadvantages, including corrosion of metal surfaces and irritation [22]. Glutaraldehyde-based disinfectants are frequently used and recommended for instrument disinfection [21]. The literature describes glutaraldehyde's high antibacterial activity, and its efficiency varies with the exposure time [24]. However, glutaraldehyde-based solutions should be used for an appropriate time due to the reported toxicity with prolonged immersion [18].

Because acrylic resins are hydrophilic, they can absorb solvents and water, triggering hydrolysis and the emergence of acrylic regions with unique optical characteristics when the absorbed liquids diffuse into the polymer network [6,34]. DBR immersion in glutaraldehyde and Micro 10+ did not affect the flexural strength, while sodium hypochlorite decreased it. This is consistent with previous studies [40,41] reporting a decrease in the flexural strength of DBRs after immersion in 1% NaOCl. This decline is due to the sorption of the NaOCl aqueous solution and its active chlorine content. The absorbed solution acts as a plasticizing agent, which could be pivotal in altering the chemical structure [42]. Moreover, the pendant monomer solubility increased due to its active chlorine, which facilitated the increase in the leachability of the remaining monomer. This was replaced with greater solution sorption [43], which is regarded as the key factor controlling the strength and surface integrity of DBRs [1].

When comparing the PMMA and 3D-printed resins based on the immersion solution, the 3D-printed resin showed low strength. This is consistent with Prpic' et al. [3], who reported a high flexural strength of conventional compared with 3D-printed DBRs. The decreased strength of 3D-printed resins may be attributed to the painting method (layer-by-layer), with apparent weak bonding between successive printed layers [3,5,14,44]. In addition to the direction of the printed layer in relation to the direction of the load applied, which was parallel to the layer direction when the specimen was vertically printed [14], another reason was attributed to the degree of conversion of photo-polymerized 3D-printed resins, as reported in a previous study, compared with conventional PMMA. As the degree of conversion decreased, the residual monomer increased, adversely affecting the strength of the printed object [3,5,44,45]. Additionally, the poor strength of the 3D-printed resins after disinfectant immersion was exaggerated due to the chemical composition of the disinfectant [46]. With immersion in solution, the resin absorbs water and the absorbed water acts as a plasticizer, affecting the mechanical properties [14,44]. A previous study [6] compared the water sorption of 3D-printed DBRs with conventional ones and reported an increase in water sorption with 3D-printed DBRs. This could also explain why the flexural strength decreased after immersion in disinfectant solution.

According to ISO-20795-1:2013 [12], a flexural strength above 65 MPa and an elastic modulus above 2000 MPa are clinically acceptable. In this context, the elastic modulus was found to be a measure of the resins' rigidity or flexibility; a higher elastic modulus denotes a more rigid material [17]. To distribute the load evenly and reduce the danger of breakage, denture base materials should have flexibility and rigidity to withstand stresses [27]. In this case, the flexural strength was mirrored by the resins' elastic moduli, with the heat-polymerized (HP) resin having a higher elastic modulus than the 3D-printed resins. This outcome is consistent with the findings of Fouda et al. [46], who examined the elastic modulus of HP and 3D-printed resins and found that the former had a lower elastic modulus. With effervescent pills, the HP and 3D-printed resins' elastic moduli dropped; however, with NaOCl, they dropped precipitously. The water and chemical uptake during submersion in these disinfectant solutions may explain this observation. Comparisons with earlier research are problematic, since no studies have assessed the impact of denture cleaners on 3D-printed resins' elastic moduli. However, the low modulus may be interpreted similarly to the flexural strength because both were tested under the same load, direction, and conditions and were reported as flexural characteristics [34].

A material's hardness is a crucial characteristic. A DBR's surface hardness indicates how much the pressures involved in mastication can be resisted. Although hardness is evaluated in numerous ways, the most practical means to determine a material's hardness is measuring its resistance to indentation [47]. A logical definition of hardness might then be the "resistance of a material to indentation"; thus, greater indentation indicates softer material [47]. It has been reported that resin immersion in disinfectant solutions weakens the secondary bond between the polymeric chains of the acrylic resins [47]. Shen et al. [35] discovered that all the resins they tested had a soft surface after being subjected to a glutaraldehyde alkaline disinfection solution with a phenolic buffer for at least 2 h. In a subsequent study [48], a surprising change in hardness was discovered after immersing specimens in glutaraldehyde for 7 days. Ma et al. [49] discovered that a phenolic-based disinfectant induced surface weakening of resins after 30 min of immersion.

The hardnesses of both resins were significantly decreased during immersion in sodium hypochlorite. In addition to the effect of disinfectants on the surface, the absorbed fluid with effective components penetrated the polymeric chains, altering the bonding and acting as a plasticizer, affecting the mechanical properties of resins and material deformation under mechanical testing [1,26,41]. The negative effect of sodium hypochlorite on both denture resins is consistent with previous studies [26,41,50] reporting a significant decrease in the hardness of 3D-printed and heat-polymerized resins after immersion in sodium hypochlorite. This is consistent with Asad et al. [48], who discovered a decrease in hardness. During the polymerization step, various amounts of residual monomer persist in the acrylic resin [51] and may function as a plasticizer, reducing the mechanical characteristics of polymerized resins [52–54]. Simultaneously, acrylic resins absorb water molecules [52,55]. Likewise, the remaining monomer can progressively leak into storage solutions, reducing the hardness of acrylic resins [23,53] that additionally serve as plasticizers, diminishing the mechanical strength of DBRs [56]. Von Fraunhofer and Suchatlampong [57] investigated the indentation resistance of denture base polymers and discovered that storage in water caused mellowing of the surface in heat-polymerized acrylic resins.

Regarding roughness analysis after immersion in solutions, the current study showed no significant difference in surface roughness between the tested specimens compared to the control group. This is consistent with a review [58] of the effect of disinfectants on the roughness of DBRs, and most of the reviewed studies reported the same findings with different disinfectant solutions. Moreover, Fotovat et al. [34] found that disinfectants produced no change in the surface roughness of 3D-printed resins. Shen et al. [35] found that glutaraldehyde-based disinfectants caused no apparent surface alteration with the standard alkaline formulation.

The present study found statistically significant differences in surface roughness between the heat-polymerized and 3D-printed acrylic for all disinfection methods. The increased roughness of the 3D-printed resins may be attributed to the orientation of the printed layers producing a stepwise effect, being perpendicular to the profilometer scan. With thermal cycling, the temperature accelerated water sorption and increased water absorption. The absorbed water moved the layers apart, impacting the surface irregularity [15,44,59]. The type and curing method of acrylic resin have been consistently reported to significantly influence surface changes after chemical disinfection because component elution may directly impact Ra [58]. Photo-polymerized resins produce more elution than heat-polymerized resins. Heat-polymerized materials have higher monomer-to-polymer conversion rates and a lower residual monomer content [60,61]. The clinical threshold for microbial adhesion is 0.2 µm, and microbial adhesion increased above this threshold [44,62]. The PMMA values were less than the clinical threshold, while all 3D-printed DBRs exceeded this value.

Fotovat et al. compared conventional and 3D-printed resins after immersion in different disinfectants and reported a significant difference in color change between resins. They attributed that change to the difference in composition (filler contents) and printing technologies (layer-by-layer) [63]. Within the category of 3D-printed resins, this component may be pivotal in the observed rise in color change [34]. This finding is consistent with our finding that no significant difference existed between the resins based on disinfectant immersion. This difference in variation could be due to the immersion time and color calculation method, as well as the resin materials used. The change in color of both resins was significantly altered with immersion in sodium hypochlorite compared to glutaraldehyde and Micro 10+. The color change with sodium hypochlorite was consistent with Carvalho et al. [64] and Rocha et al. [25], who found a significant increase in the change in color of PMMA after immersion in sodium hypochlorite. This was consistent with previous studies [19,50,65] confirming the pronounced color change with sodium hypochlorite over other disinfectants. The color changes were attributed to disinfectant solvents permeating the polymer network, expanding the intermolecular spaces. Subsequently, this process resulted in exchanges of internal and external pigments, alterations in the polymer matrix, and the chemical degradation and dissolution of their compounds, leading to color changes [30,63,65]. A previous study [66] reported that sodium hypochlorite exhibited whitening of resins through the oxidation of resin surfaces, consistent with the finding of the present study. Conversely, another study [67] disagreed with the present study, finding that the color steadiness of DBRs was enhanced following immersion in 2% alkaline glutaraldehyde as well as 0.5% sodium hypochlorite solution. These variations in results are attributed to differences in material type and immersion time.

Despite the color changes with sodium hypochlorite, all changes fell within the clinically acceptable value for color changes based on the NBS value. Regarding the thresholds established in the study by Fotovat et al. [34], the color change observed in all three groups was deemed clinically acceptable. This may be due to the immersion time compared to other studies reporting noticeable color changes after immersion for a prolonged time [26,34,68]. Finally, the conventional denture bases outperformed the 3D-printed bases in color stability. Specifically, sodium hypochlorite caused substantially more color change than the other disinfectants. Among all groups, the conventional DBR samples immersed in glutaraldehyde and Micro 10+ showed the least color change.

Regarding disinfectant solutions, regardless of material type, sodium hypochlorite adversely affected the tested properties. Although sodium hypochlorite showed a strong antimicrobial effect, other disinfectants could be recommended as alternatives, as these showed antimicrobial effectiveness without adverse effects on the strength, surface, and color properties. Regarding resin type, although PMMA showed superior performances to the 3D-printed DBRs, all values were within the clinically acceptable range except the roughness of the 3D-printed resins, which exceeded the clinical threshold. Clinically, glutaraldehyde and Micro 10+ could be recommended as disinfectants in dental laboratories and dental offices to control infections and overcome cross-contamination.

The use of bar- and disk-shaped specimens rather than a denture configuration and the duration of immersion for each disinfectant solution are the two limitations of this study. Other limitations are this study's in vitro nature, the restriction to specimens fabricated in the laboratory, and the absence of aged specimens returned to the laboratory for repair or modification after patient use for a prolonged time. Therefore, different disinfectant effects on a real denture base with different immersion times should be investigated in future studies. Additionally, aging specimens using thermal cycling in chewing simulators representing the specimens' return from the clinic to the laboratory for repair or adjustment are required in future investigations.

5. Conclusions

The disinfectant type affects the properties of DBRs. Sodium hypochlorite has adverse effects on the flexural strength, elastic modulus, hardness, and change in color of PMMA and 3D-printed DBRs. Conversely, other disinfectants do not affect the tested properties. Based on the findings of the present study, glutaraldehyde and Micro 10+ could be recommended for DBR disinfection to overcome cross-contamination between dental laboratories and dental offices.

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