Heliyon 9 (2023) e15314

Contents lists available at ScienceDirect

Heliyon

journal homepage: www.cell.com/heliyon

Research article

Mechanical properties of CAD/CAM polylactic acid as a material for interim restoration

Won-Il Choi [a], Lee-gang Yoo [a], Yu-ri Kim [a], Bock-Young Jung [a,*]

[a] Department of Advanced General Dentistry, College of Dentistry, Yonsei University, Seoul, South Korea

ARTICLE INFO

Keywords:
Polylactic acid
Flexural strength
CAD/CAM
Dental polymer
Mechanical properties

ABSTRACT

Statement of problem: Biomaterials, including polymethyl methacrylate (PMMA) and bisacrylate, have been widely used as conventional interim materials and may exhibit cytotoxicity or systemic toxicity.

Purpose: This study was designed to compare the mechanical properties of polylactic acid (PLA) as an alternative to conventional dental polymers for computer-aided design and manufacturing (CAD/CAM).

Material and methods: Four groups (n = 20 per group) of CAD/CAM polymers were assessed. Specimens of PLA (PLA Mill) and PMMA (PMMA Mill) for subtractive manufacturing, PLA for fused deposition modeling (PLA FDM), and bisphenol for additive manufacturing by stereolithography (Bisphenol SLA) were fabricated into 2-mm-wide, 2-mm-thick and 25-mm-long specimens using a milling machine, an FDM printer, and an SLA printer, respectively.

The flexural strength (FS) and elastic modulus (EM) were calculated. The surface roughness and Shore D hardness were analyzed with a 3D optical surface roughness analyzer and a Shore durometer, respectively.

Results: PLA Mill showed the lowest FS (64.9 ± 8.28), followed by PLA FDM (104.27 ± 4.42 MPa), PMMA Mill (139.2 ± 20.95 MPa), and Bisphenol SLA (171.56 ± 15.38 MPa), with statistically significant differences. PLA FDM showed the highest EM, followed by PLA Mill, Bisphenol SLA, and PMMA Mill. Significant differences were observed not only between PMMA Mill and Bisphenol SLA but also between PLA FDM and PLA Mill. The lowest Shore D hardness was observed for PLA FDM, followed by PLA Mill, PMMA Mill, and Bisphenol SLA, which showed the highest value among the 4 groups, with significance. The highest values for the surface roughness parameters were observed for PLA Mill, and the lowest were observed for Bisphenol SLA.

Conclusions: Among the tested CAD/CAM polymers, Bisphenol SLA was the most durable material, and the mechanical properties of PLA FDM were within the clinically acceptable range.

1. Introduction

Environmental and safety issues have recently drawn attention and led scientists and manufacturers to investigate new nontoxic or biodegradable materials, which is a global trend in industry as well as in the biomedical field.

Biomaterials, including polymethyl methacrylate (PMMA) and bisacrylate, have been widely used as conventional interim

* Corresponding author.
E-mail address: jby1004@yuhs.ac (B.-Y. Jung).

https://doi.org/10.1016/j.heliyon.2023.e15314
Received 19 December 2022; Received in revised form 20 March 2023; Accepted 3 April 2023
Available online 9 April 2023

Abbreviations

PLA	polylactic acid
EM	elastic modulus
FS	flexural strength
AM	additive manufacturing
SM	subtractive manufacturing

materials due to their mechanical properties, physiochemical properties and moldability [1]. Several in vitro studies on the biocompatibility of PMMA resin have reported that residual monomers, such as methyl methacrylate, formaldehyde, biphenyl and phenyl benzoate, may exhibit cytotoxicity or systemic toxicity [2–5]. The concentrations of these residual monomers and the eluent additives of PMMA depend on several parameters, including the polymerization cycle, polymerization method, and post-polymerization treatment [6–10].

Among biopolymers, polylactic acid (PLA) is one of most widely used polymers due to its numerous advantages. PLA is derived from nontoxic renewable sources (e.g., corn, wheat, sugar, and other agricultural crops) in an ecofriendly manner that consumes carbon dioxide in air. The most interesting feature of PLA is its biocompatibility, especially in biomedical applications, and PLA has been approved by the U.S. Food and Drug Association (FDA) for direct contact with biological fluids [11]. After approval by the FDA, medical applications of PLA have expanded to medical implants, orthopedic devices, covering membranes, tissue engineering scaffolds and dermatological sutures, and investigations on the compatibility of PLA with other materials through proper surface modification or blend preparation for potential dentistry applications [11].

Furthermore, PLA produces no toxic effects, and its degradation byproducts do not interfere with tissue healing, as PLA hydrolyzes into an alpha-hydroxyl acid in the human body, which is incorporated into the tricarboxylic acid cycle and then excreted as water and carbon dioxide [12–15]. In addition, PLA has better thermal processability for film production via extrusion, injection molding, blow molding, fiber spinning, and thermoforming [14]. Regarding energy use, PLA production requires 25–55% less energy than other polymer production methods [12].

Even though the chemical, mechanical and biological properties of PLA are considered to be comparable to those of conventional polymers, PLA utilization is limited due to the need for chemical and physical modification to address the toughness, degradation rate, hydrophobicity, and lack of reactive side-chain groups. Although the tensile strength and elastic modulus (EM) of PLA are comparable to those of other polymers, its poor toughness limits the use of PLA when plastic deformation is expected at sites under higher stress.

The mechanical properties, such as the flexural strength (FS), EM, and elongation at break, and crystallinity of PLA are dependent on the molecular weight (Mw) and stereochemical composition of the PLA backbone [15]; when the Mw increased from 50 to 150 to 200 kDa, the tensile strength increased from 15.5 to 80 to 150, respectively [16]. The stereochemical composition is controlled by the polymerization of D-lactide, L-lactide, and DL-lactide and affects the mechanical properties of PLA. Previous studies reported that, depending on the stereochemistry, the FS and EM varied in wide ranges of 100–119 MPa and 3750–4150 MPa, with the lowest FS and highest EM reported for PDLLA polymerization [15,16].

With recent advancements in dental CAD/CAM technologies, both subtractive manufacturing (SM) and additive manufacturing (AM) have been actively utilized to fabricate patient-specific customized devices, including interim restorations, surgical stents, and rapid prototype skulls. Compared to SM, which is still more frequently used, AM has considerable advantages for manufacturing dental devices or prosthetic restorations because it saves raw materials, energy, rotatory tools, and time [17].

Fused deposition modeling (FDM), stereolithography (SLA), and digital light processing (DLP) are typical AM methods used in the dental field. The resolution of DLP and SLA products is known to be higher than that of FDM, and SLA is superior to DLP in terms of surface texture, but DLP is faster than SLA [18,19]. The operational variables of 3D printing technology, including the printing angle and direction, light source speed and intensity, irradiation method and layer thickness, can affect the surface roughness, dimensional accuracy, and strengths of the products [20–22].

A number of CAD/CAM dental polymers have been developed, and PMMA for SM, bisphenol resin for AM, and polyaryletherketone (PAEK) for both AM and SM have been widely used in dentistry. However, their inherent limitations inhibit their use in clinical applications; for example, PMMA for 3D printing has a high shrinkage rate during light curing and exhibits poor mechanical properties, and the Mw and viscosity of bisphenol A-glycidyl methacrylate (Bis-GMA) and urethane dimethacrylate (UDMA) are high, which reduces the fluidity for 3D printing [23].

The FS of PLA was known to be 108 MPa based on tests conducted according to ISO 10477 [12,24,25]. A prior study investigated the mechanical and fit properties of crowns milled from PLA compared to those of PMMA and PEEK and reported that the most durable material in terms of compressive loading was PEEK (840.90 $\pm$ 13.23 N), followed by PMMA (733.30 $\pm$ 9.00 N), while the least durable material was PLA (644.50 $\pm$ 10.79 N) [1].

Despite the advantages of PLA, there have been few direct comparisons of its mechanical and biophysical properties with those of conventional CAD/CAM materials for interim dental prostheses performed under controlled experimental conditions. This study was designed to compare the FS, EM, Shore D hardness and surface roughness to evaluate the potential use of PLA as an alternative for CAD/CAM products.

2. Material and methods

Prior to this in vitro study, PLA was approved as a material for interim crowns and bridges by the Korea Food and Drug Association (KFDA) after a series of tests, such as skin sensitization studies, intracutaneous reactivity studies, oral mucosa irritation studies, in vitro cytotoxicity tests, and dental device tests, which were performed by the Yonsei University Medical Center.

2.1. Test specimens and materials

Four types of conventional CAD/CAM polymers used for interim restorations were tested. PLA Mill and PMMA Mill specimens were fabricated with a milling machine, PLA FDM was fabricated using an FDM printer, and Bisphenol SLA was fabricated using an SLA printer (Table 1). Twenty specimens for each group were prepared by two technicians according to the customized preset parameters for each CAD/CAM device. The specimens were virtually designed with a CAD software program (exocad GmbH), converted into stereolithographic (STL) format, and then milled or 3D printed into a rectangular bar with a width of 2 mm, thickness of 2 mm and length of 25 mm, in accordance with ISO 10477 [25]. The Bisphenol SLA specimens were immersed in 100% isopropyl alcohol to remove resin monomers (Medifive, Tornado), and postpolymerization was performed for 210 s by using a UV light-polymerization unit (LC-3D print box, NextDent); however, specimens in the FDM group did not undergo postpolymerization processing. The dimensions of all specimens were checked with a digital caliper. The sample sizes for the study groups were determined via a sensitivity power analysis with 80% power, a 5% significance level [1], and an effect size of 0.4 [26] using a software program (G*Power, v3.1.9.2; Heinrich-Heine-Universität Düsseldorf).

2.2. Three-point FS tests

A three-point FS test was conducted with a universal testing machine (Instron 3366; Instron Corporation). The specimens were located on a holding jig with a support span of 20 mm and the load pressure was applied on the center of the specimen with a 0.5 kN load cell at a cross head speed of 1.0 mm/min until the specimen fractured according to ISO 10477 [25]. The FS and EM values were calculated by transforming the maximum fracture value recorded in Newtons (N) with the following formulas:

for FS, $\sigma_f = (3 \times P \times L)/(2 \times b \times h^2)$

for EM, $E_f = (L^3 \times m)/(4 \times b \times h^3)$

P: maximal load (N), L: support span (mm), b: width of the specimen at the failure site (mm), h: height of the specimen at the failure site, and m: gradient of the initial straight-line portion of the load–deflection curve (N/mm).

2.3. Shore D hardness

Five measurements were performed at 25 °C for each specimen according to ISO 868 [27] by placing the specimen under the indenter of a Shore durometer (HPSD; Schmidt), and the mean value was recorded.

2.4. Surface roughness

The surfaces of two specimens from each group were analyzed using a 3D optical surface roughness analyzer with a vertical resolution of 0.05 nm and a root mean square (RMS) repeatability of 0.01 nm (Contour GT-X3 BASE; Bruker) at 9 locations per specimen. The centerline average roughness (Ra) and ten-point median height (Rz) values were calculated. The objective magnification was 50 × , and the zoom was 2 × . The size of the field view was 0.09 × 0.066 mm^2.

Table 1
List of materials studied.

Group	Polymer	Manufacturer	Manufacturing method	Manufacturer
PLA FDM	Polylactic acid	CUBICON Style –Plus-A15D CUBICON CO Korea	Fused deposition modeling Φ = 1.75 mm, Layer height: 0.1 mm Nozzle: 0.4 mm Speed: 60 m/s	KPLA, 3D KOREA CO Korea
PLA Mill	Polylactic acid	R2 solution MEGAGEN Korea	Subtractive milling	KPLA, 3D KOREA CO Korea
PMMA Mill	Poly methyl methacrylate	R2 solution MEGAGEN Korea	Subtractive milling	VIPI BLOCK TRILUX®, Vipi Odonto Products
Bisphenol SLA	Bis-A ethoxylate dimethacrylate	KARV LP 550 Shinwod Dental Korea	Stereolithography Layer height: 0.1 mm, Curing time: 7 s	ODS C&B ODS CO Korea

2.5. Scanning electron microscopy (SEM)

One specimen per group was observed with field emission SEM (FE-SEM) (JEOL-7800F; JEOL, Ltd.) at an acceleration voltage of 5 kV, except PLA specimens were observed at an acceleration voltage of 10 kV. The specimen from each material group was left to dry at room temperature for 24 h and then sputter-coated with gold and palladium for 180 s before FE-SEM examination.

2.6. Statistical analysis

When Shapiro–Wilk and Leven's tests indicated normality and homogeneity of variance, one-way ANOVA and Tukey's post hoc test were performed; the nonparametric Kruskal–Wallis and Mann–Whitney tests were used to analyze the data that did not show normality. The Mann–Whitney test with the Bonferroni correction for multiple comparisons was performed to compare individual group. All statistical analyses were performed using the SPSS Statistics package (IBM SPSS; IBM Corp) and reviewed by an independent statistician.

3. Results

The means and standard deviations of the FS, EM, Shore D hardness and surface roughness values for each group are provided in Table 2 and Table 3.

3.1. FS and EM

The Kruskal–Wallis test demonstrated significant differences in the FS ($p = .000$) and EM ($p = .003$) values among all groups. The Mann–Whitney test with the Bonferroni correction for multiple comparisons showed that for the FS, there were significant differences among the tested groups, while for the EM, there were significant differences between the Bisphenol SLA and PMMA Mill groups as well as between the PLA FDM and PLA Mill groups.

The PLA Mill group showed the lowest FS, followed by the PLA FDM, PMMA Mill, and Bisphenol SLA groups; moreover, PLA FDM showed the highest EM, followed by PLA Mill, Bisphenol SLA, and PMMA Mill, as shown in Figs. 1 and 2.

3.2. Shore D hardness and surface roughness

The Shapiro–Wilk test indicated normality and homogeneity of variance; therefore, one-way ANOVA and Tukey's post hoc test were performed. The lowest Shore D hardness was observed for PLA FDM, followed by PLA Mill, PMMA Mill, and Bisphenol SLA, which showed the highest value among all groups, with significant differences (Table 3, Fig. 3). Total surface roughness was represented using the color variation between two main colors: red and blue. From the mean line on the height contour, red represented the peaks of height and blue indicated the valleys' depth. The values on the vertical axes of the parameter and color variation between red and blue represented the overall surface roughness. In consideration of the Ra value of each group, the image with the most similar Ra value was selected among the various images of each group (Fig. 4a–d).

The highest Ra and Rz values were observed for the PLA Mill group, and the Bisphenol SLA group exhibited the lowest values. There was no significant difference between the PLA Mill and PMMA Mill groups in terms of the Ra value, while there were overall significant differences in the Rz parameter among the 4 different groups (Table 3, Fig. 4a–d).

3.3. FE-SEM

Representative FE-SEM images showed that the surfaces of PLA FDM and Bisphenol SLA were smoother than those of PLA Mill and PMMA Mill (Fig. 5a–d, Fig. 6a–e).

A few sites of unpolymerized PMMA were observed at 2000 × magnification (Fig. 6c), and there were some pore-like defects in the PLA Mill group (Fig. 5b). The FE-SEM results agree with the surface roughness results (Fig. 4a–d).

4. Discussion

We characterized the mechanical properties of PLA in comparison to those of conventional CAD/CAM polymers as a preliminary

Table 2
Flexural strength and elastic modulus (mean ± SD).

Group	Flexural strength (MPa)	Elastic modulus (N/㎟)
PLA FDM	104.27 ± 4.42[a]	6660.56 ± 5354.67[ab]
PLA Mill	64.9 ± 8.28[b]	4995.1 ± 4564.75[cd]
PMMA Mill	139.2 ± 20.95[c]	3243 ± 396.7[bd]
Bisphenol SLA	171.56 ± 15.38[d]	4548.33 ± 4221.89[ac]

Different letters in the columns indicate statistically significant differences ($p < .0083$).

Table 3
Shore D hardness and surface roughness (mean ± SD).

Material	Shore D (HSD)	Ra (μm)	Rz (μm)
PLA FDM	73.2 ± 1.19[a]	0.24 ± 0.23[a]	6.25 ± 2.8[a]
PLA Mill	80.20 ± 0.47[b]	1.25 ± 0.32[b]	16.2 ± 4.83[b]
PMMA Mill	87.72 ± 0.36[c]	1.16 ± 0.27[b]	10.93 ± 2.7[c]
Bisphenol SLA	90.6 ± 0.45[d]	0.18 ± 0.07[a]	3.24 ± 0.85[d]

Different letters in the columns indicate statistically significant differences (α = 0.05).

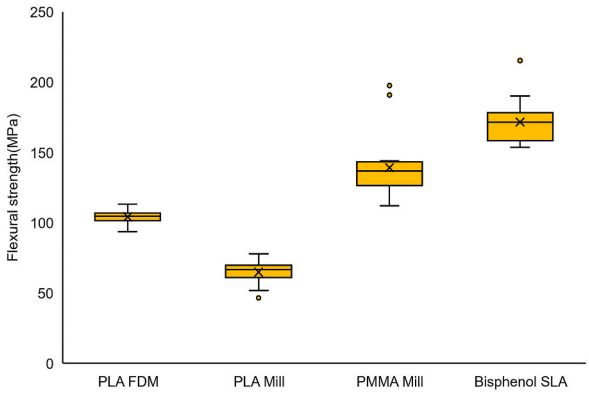

Fig. 1. Flexural strength of test groups. The boxplot diagrams show the mean value (x), outliers (°), median value (bar), quartile values (box), and minimum and maximum values (whiskers).

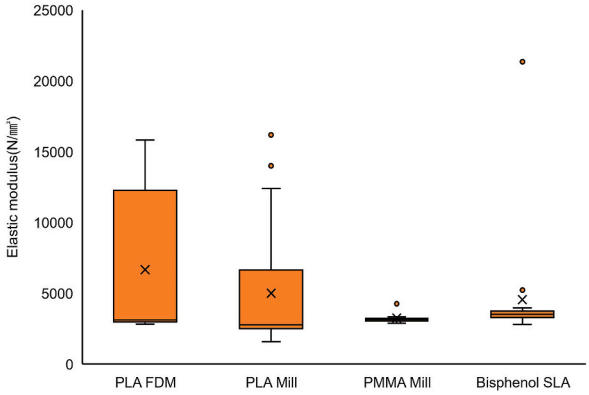

Fig. 2. Elastic modulus of test groups. The boxplot diagrams show the mean value (x), outliers (°), median value (bar), quartile values (box), and minimum and maximum values (whiskers).

study prior to any clinical investigations. The null hypothesis that there are no significant differences in the mechanical properties among the different test groups was rejected, as statistically significant differences were found among the test groups.

A three-point FS test was performed based on ISO 10477 because the PLA used in this study could not withstand the biaxial FS test [28]. For materials with high toughness, the biaxial FS test is considered a more reliable method than the three-point FS test since in the biaxial test, radial stress and tangential stress are expressed to accurately measure the strength without considering edge failure [29, 30].

The FS of PLA FDM (104.27 ± 4.42) was significantly lower than those of PMMA Mill (139.2 ± 20.95) and Bisphenol SLA (171.56 ± 15.38) and significantly higher than that of PLA Mill (64.9 ± 8.28). The FS of all specimens was over 65 MPa, which has been reported as the minimum FS suitable for polymeric materials used in dental applications; additionally, the FS of PLA FDM in this study was similar to the value of 108 MPa reported in a previous study [24], but Farah et al. reported differences in FS depending on stereochemistry ranging from 88 to 119 MPa [12].

This study was also intended to compare the mechanical properties of PLA specimens manufactured using AM and SM technologies. Martin et al. reported that a milled specimen showed a significantly higher fracture resistance than the printed specimen using commercial products [31]. However, in this study, the FS of PLA Mill was significantly lower than that of PLA FDM. Why is the FS of

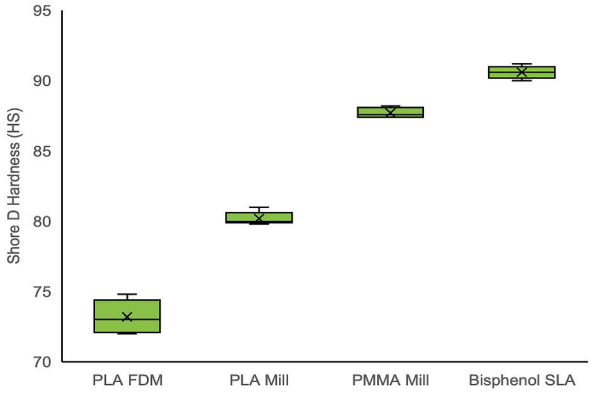

Fig. 3. Shore D hardness of test groups. The boxplot diagrams show the mean value (x), median value (bar), quartile values (box), and minimum and maximum values (whiskers).

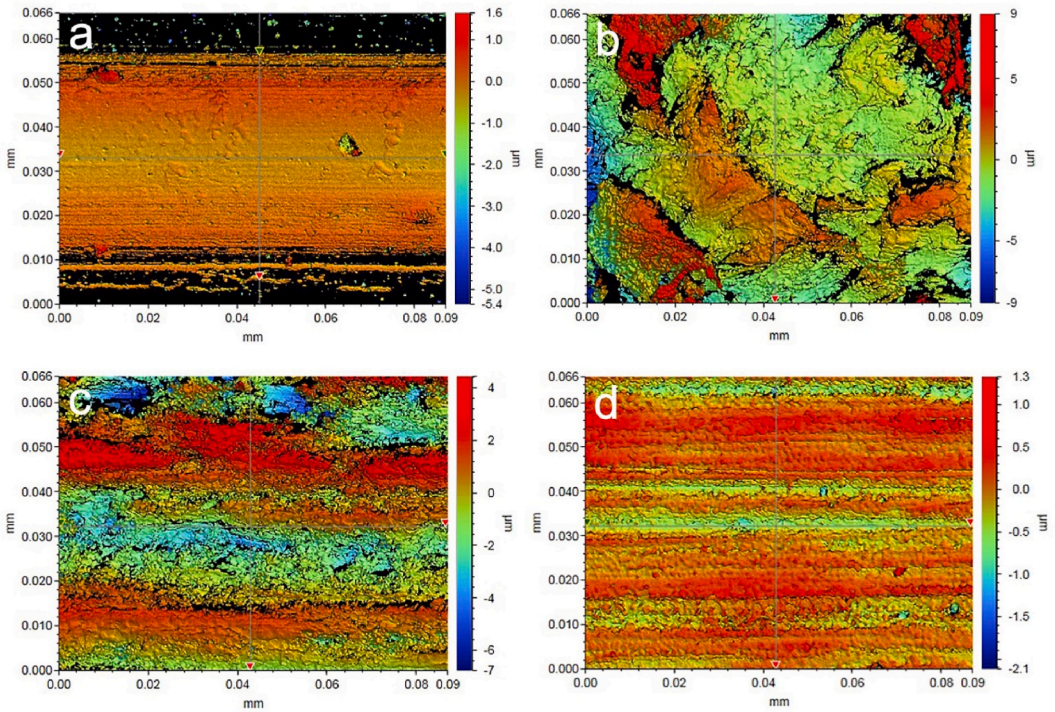

Fig. 4. Surface roughness images of test groups: (a) PLA FDM; (b) PLA Mill; (c) PMMA Mill; and (d) Bisphenol SLA. (Objective magnification: 50 ×, Zoom: 2 ×, Size of the field view: 0.09 × 0.066 mm^2).

PLA Mill lower than that of PLA FDM even though identical powder was used in both groups? The fact that the PLA Mill blocks were additively printed with no angle and the PLA Mill specimens were fabricated using wet milling technology could explain this result. In general, a number of parameters associated with AM, including layer height, extrusion temperature, bed temperature, feed rate, annealing, void inclusion, nozzle size and raster orientation, could affect the mechanical behaviors of additively manufactured products [1,30,31]. Merve et al. assessed the fit and FS of provisional crowns manufactured by milling an additively manufactured disc of L-PLA and indicated that PLA may be a good alternative to provisional materials; it showed a fracture strength of 664.50 N, which was greater than the average occlusal force of 250–350 N [1].

The Shore D hardness of PLA Mill (80.20 ± 0.47 HSD) was similar to the previously reported value of 85 HSD [24] and significantly higher than the value for PLA FDM (73.2 ± 1.19 HSD). This result was contrary to the result of the FS.

Printing conditions affect the molecular alignment, crystallinity, and postprocessing annealing of printed polymers, resulting in a variety of mechanical properties [32]. In this study, the nozzle temperature and bed temperature were kept at approximately 200 °C and 65 °C, respectively, to maximize the balance between the degree of crystallinity and postprocessing annealing. Additionally,

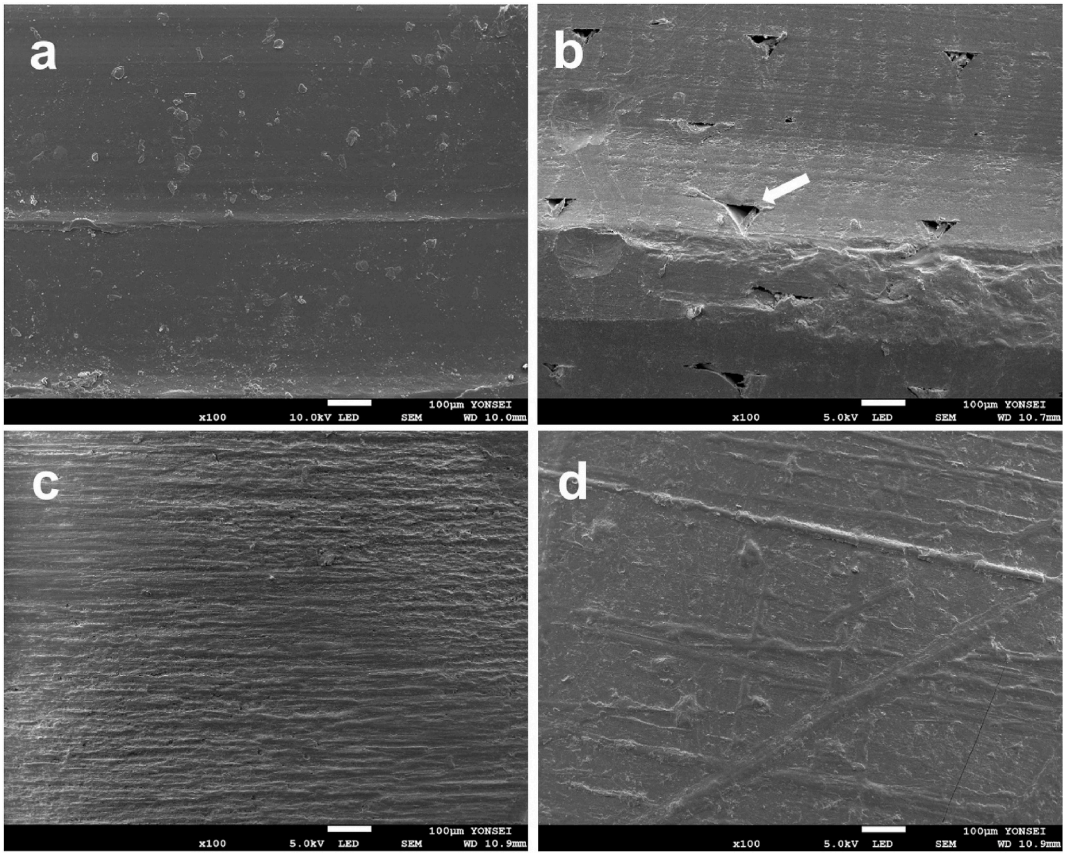

Fig. 5. Scanning electron microscopy (FE-SEM) images of test groups at 100 × magnification: (a) PLA FDM; (b) PLA Mill; (c) PMMA Mill; and (d) Bisphenol SLA.

postprocessing annealing is the main process for improving mechanical properties; hence, the bed temperature is usually kept above the glass-transition temperature (T_g, approximately 60 °C) to maximize bonding between the deposited layers [32].

A 135-degree printing orientation for PLA FDM was consistently used in this study based on a previous study reporting that this angle was the ideal orientation for producing optimal mechanical properties [32]. Alharbi et al. concluded that a specimen printed perpendicular to the load direction had higher compressive strength [20], and Osman et al. recommended 135° for DLP [33]. Another previous study reported that printing at 30° resulted in the highest FS [17].

This study evaluated the surface of a specimen from each group by FE-SEM and surface roughness analysis. A previous study on 3D-printed resin material reported that each layer was thick and that the melted filaments were thinly stretched in the last layer in the FDM group, as observed at 200 × magnification [17]. However, in this study, FE-SEM images of PLA FDM at magnifications of 100 × or more showed a smooth surface with irregularly distributed, small, round grains (Fig. 5a and 6a) instead of piled-up filaments. In the PMMA Mill group (Fig. 5c), the images showed a rougher surface than observed in the PLA FDM and Bisphenol SLA groups (Fig. 5a and d). Oyar et al. reported results similar to those of this study regarding the SEM features and surface roughness of milled PMMA [34]. In this study, some cracks and a few sites of nonpolymerized PMMA powder were observed (Fig. 6c and d), but a previous study showed a homogeneous image for a milled PMMA resin specimen [17].

In this study, the results of the surface roughness and FE-SEM analyses were in agreement and mutually supportive. Generally, the specimens in the SLA group were cleaner and smoother because the interlayer transition was well filled with cured resin material [17]. The surfaces of the Bisphenol SLA specimens were homogenous at 100 × magnification (Fig. 5d), but some waveform irregularities and valleys were observed at 10,000 × magnification (Fig. 6e). Additionally, some pore-like spaces on the surface of PLA Mill were observed at 100 × magnification (Fig. 5b), which might have been caused by an incomplete interlayer transition due to the nozzle and/or bed temperature control failure. It can be inferred that if a PLA block is formed under compressive pressure, an increased FS and no internal pore-like defects can be expected.

One of the main issues is how stable an interim device made of PLA material is over time in a wet atmosphere such as the oral cavity. PLA degrades through the hydrolysis of backbone ester groups, and the degradation rate depends on the crystallinity, molecular weight and distribution, morphology, water diffusion rate and stereoisomeric content of the PLA.

Moisture diffuses into a polymer to form pores and weaken the microstructure; thereafter, biodegradation proceeds via increased moisture absorption in the wet atmosphere. As PLA is a hydrophobic and aliphatic polyester, the initial rate of hydrolysis is very slow at

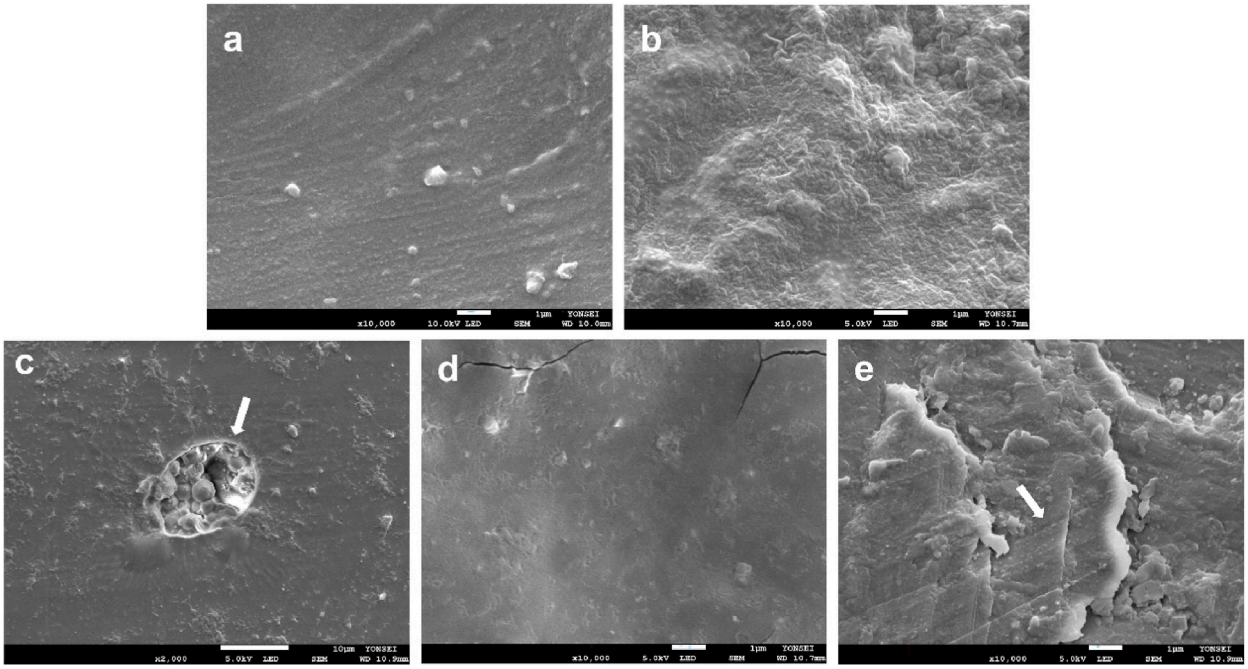

Fig. 6. Scanning electron microscopy (FE-SEM) images of test groups: (a) PLA FDM at 10,000 × magnification; (b) PLA Mill at 10,000 × magnification; (c) PMMA Mill at 2000 × magnification; (d) PMMA Mill at 10,000 × magnification; and (e) Bisphenol SLA at 10,000 × magnification.

the end of the polymer chain. It was reported that more than 90% remained after 133 days at 37 °C and after 28 days at 60 °C; furthermore, if a carboxyl group is formed at the end of the chain to form a water-soluble oligomer, the hydrolysis accelerates rapidly [12]. In general, it is considered that a material used for an interim prosthesis could contact tissue, bone and dentin both internally and externally over 24 h for 30 days according to ISO 7405. Therefore, it can be inferred that the mechanical properties of PLA as an interim prosthesis in the oral cavity can be stably maintained within 30 days.

This was not only an in vitro study but also a type of preliminary study for clinical application. Further research on PLA as a promising dental polymer should include methods to improve mechanical strength through stereochemical modifications and polymerization methods. In addition, since the mechanical properties of dental polymers are adversely affected by the absorption of moisture in the oral cavity, it is necessary to investigate the effect of water absorption, pH and temperature on the mechanical properties.

5. Conclusion

Within the limitations of this in vitro study, it can be stated that the most durable material for CAD/CAM provisional restoration is bisphenol resin and that the PLA could be used as an alternative to conventional provisional restoration materials.

Declaration of competing interest

The authors declare that there are no conflicts of interest regarding the publication of this study.

Author contribution statement

Won-Il Choi: Performed the experiments; Contributed reagents, materials, analysis tools or data; Wrote the paper. Lee-gang Yoo: Analyzed and interpreted the data; Contributed reagents, materials, analysis tools or data. Yu-ri Kim: Performed the experiments; Analyzed and interpreted the data. Bock-Young Jung: Conceived and designed the experiments; Wrote the paper.

Funding statement

This work was supported by the National Research Foundation of Korea (NRF) grant funded by the Korean government (MSIT) [grant number 2020R1A2C2004893].

Data availability statement

Data will be made available on request.

Declaration of interest's statement

The authors declare that they have no competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Appendix A. Supplementary data

Supplementary data related to this article can be found at https://doi.org/10.1016/j.heliyon.2023.e15314.

References

[1] M. Benli, B. Eker-Gümüş, Y. Kahraman, O. Huck, M. Özcan, Can polylactic acid be a CAD/CAM material for provisional crown restorations in terms of fit and fracture strength? Dent. Mater. J. 40 (3) (2021) 772–780.
[2] T. Koda, H. Tsuchiya, M. Yamauchi, S. Ohtani, N. Takagi, J. Kawano, Leachability of denture-base acrylic resins in artificial saliva, Dent. Mater. 6 (1) (1990) 13–16.
[3] E.C. Munksgaard, Plasticizers in denture soft-lining materials: leaching and biodegradation, Eur. J. Oral Sci. 113 (2) (2005) 166–169.
[4] R. Brozek, R. Rogalewicz, R. Koczorowski, A. Voelkel, The influence of denture cleansers on the release of organic compounds from soft lining materials, J. Environ. Monit. 10 (6) (2008) 770–774.
[5] M. Nakamura, H. Kawahara, Long-term biocompatibility test of denture base resins in vitro, J. Prosthet. Dent 52 (5) (1984) 694–699.
[6] H. Tsuchiya, Y. Hoshino, H. Kato, N. Takagi, Flow injection analysis of formaldehyde leached from denture-base acrylic resins, J. Dent. 21 (4) (1993) 240–243.
[7] S. Baker, S.C. Brooks, D.M. Walker, The release of residual monomeric methyl methacrylate from acrylic appliances in the human mouth: an assay for monomer in saliva, J. Dent. Res. 67 (10) (1988) 1295–1299.
[8] T. Kawahara, Y. Nomura, N. Tanaka, W. Teshima, M. Okazaki, H. Shintani, Leachability of plasticizer and residual monomer from commercial temporary restorative resins, J. Dent. 32 (4) (2004) 277–283.
[9] M. Kawaguchi, Y. Takahashi, T. Fukushima, T. Habu, Effect of light-exposure duration on the amount of leachable monomers from light-activated reline material, J. Prosthet. Dent 75 (2) (1996) 183–187.
[10] A.F. Boeckler, D. Morton, S. Poser, K.E. Dette, Release of dibenzoyl peroxide from polymethyl methacrylate denture base resins: an in vitro evaluation, Dent. Mater. 24 (12) (2008) 1602–1607.
[11] M.S. Singhvi, S.S. Zinjarde, D.V. Gokhale, Polylactic acid: synthesis and biomedical applications, J. Appl. Microbiol. 127 (6) (2019) 1612–1626.
[12] S. Farah, D.G. Anderson, R. Langer, Physical and mechanical properties of PLA, and their functions in widespread applications - a comprehensive review, Adv. Drug Deliv. Rev. 107 (2016) 367–392.
[13] R.M. Rasal, A.V. Janorkar, D.E. Hirt, Poly (lactic acid) modifications, Prog. Polym. Sci. 35 (3) (2010) 338–356.
[14] R. Auras, B. Harte, S. Selke, An overview of polylactides as packaging materials, Macromol. Biosci. 4 (9) (2004) 835–864.
[15] D. Garlotta, A literature review of poly (lactic acid), J. Polym. Environ. 9 (2) (2001) 63–84.
[16] K. Van de Velde, P. Kiekens, Biopolymers: overview of several properties and consequences on their applications, Polym. Test. 21 (4) (2002) 433–442.
[17] S.M. Park, J.M. Park, S.K. Kim, S.J. Heo, J.Y. Koak, Flexural strength of 3D-printing resin materials for provisional fixed dental prostheses, Materials 13 (18) (2020).
[18] G.H. Wu, S.H. Hsu, Review: polymeric-based 3D printing for tissue engineering, J. Med. Biol. Eng. 35 (3) (2015) 285–292.
[19] T. Finnes, High definition 3d printing–comparing sla and fdm printing technologies, J. Undergrad. Chem. Res. 13 (1) (2015) 3.
[20] N. Alharbi, R. Osman, D. Wismeijer, Effects of build direction on the mechanical properties of 3D-printed complete coverage interim dental restorations, J. Prosthet. Dent 115 (6) (2016) 760–767.
[21] A. Unkovskiy, P.H. Bui, C. Schille, J. Geis-Gerstorfer, F. Huettig, S. Spintzyk, Objects build orientation, positioning, and curing influence dimensional accuracy and flexural properties of stereolithographically printed resin, Dent. Mater. 34 (12) (2018) e324–e333.
[22] A. Awada, D. Nathanson, Mechanical properties of resin-ceramic CAD/CAM restorative materials, J. Prosthet. Dent 114 (4) (2015) 587–593.
[23] H. Chen, S.Y. Lee, Y.M. Lin, Synthesis and formulation of PCL-based urethane acrylates for DLP 3D printers, Polymers 12 (7) (2020).
[24] P. Molinero-Mourelle, S. Canals, M. Gómez-Polo, M.F. Solá-Ruiz, J. Del Río Highsmith, A.C. Viñuela, Polylactic acid as a material for three-dimensional printing of provisional restorations, Int. J. Prosthodont. (IJP) 31 (4) (2018) 349–350.
[25] I.O.f. Standardization, Dentistry: Polymer-based Crown and Bridge Materials, ISO 10477 (1992) 1992 (E).
[26] J. Cohen, Statistical Power Analysis for the Behavioral Sciences, Routledge, 2013.
[27] I. ISO, 868: 2003, Plastics and Ebonite–Determination of Indentation Hardness by Means of a Durometer (Shore Hardness), International Organization for Standardization, Geneva, Switzerland, 2003.
[28] S. Ban, K.J. Anusavice, Influence of test method on failure stress of brittle dental materials, J. Dent. Res. 69 (12) (1990) 1791–1799.
[29] W.C. Wagner, T.M. Chu, Biaxial flexural strength and indentation fracture toughness of three new dental core ceramics, J. Prosthet. Dent 76 (2) (1996) 140–144.
[30] D.-G. Seo, B.-D. Roh, The comparison of relative reliability on biaxial and three point flexural strength testing methods of light curing composite resin, J. Korean Acad. Conserv. Dent. 31 (1) (2006) 58–65.
[31] N. Martín-Ortega, A. Sallorenzo, J. Casajús, A. Cervera, M. Revilla-León, M. Gómez-Polo, Fracture resistance of additive manufactured and milled implant-supported interim crowns, J. Prosthet. Dent 127 (2) (2022) 267–274.
[32] C. Benwood, A. Anstey, J. Andrzejewski, M. Misra, A.K. Mohanty, Improving the impact strength and heat resistance of 3D printed models: structure, property, and processing correlationships during fused deposition modeling (FDM) of poly(lactic acid), ACS Omega 3 (4) (2018) 4400–4411.
[33] R.B. Osman, N. Alharbi, D. Wismeijer, Build angle: does it influence the accuracy of 3D-printed dental restorations using digital light-processing technology? Int. J. Prosthodont. (IJP) 30 (2) (2017) 182–188.
[34] P. Oyar, M. Ulusoy, R. Durkan, Effects of repeated use of tungsten carbide burs on the surface roughness and contact angles of a CAD-CAM PMMA denture base resin, J. Prosthet. Dent 128 (2022) 1358–1362.