RESEARCH AND EDUCATION

Fracture resistance and 2-body wear of 3-dimensional−printed occlusal devices

Anna-Maria Lutz, Dr med dent,[a] Rüdiger Hampe, Dipl.-Ing. (FH), MSc,[b] Malgorzata Roos, PD, Dr,[c]
Nina Lümkemann, MSc,[d] Marlis Eichberger, DDT,[e] and Bogna Stawarczyk, PD, Dr Dipl.-Ing. (FH), MSc[f]

Using occlusal devices is a conservative measure to protect the teeth from excessive occlusal wear, to relax the muscles of mastication, and to ameliorate or alleviate the myogenous symptoms of temporomandibular disorders. The design of devices differs according to their intended use. One of the most common types of occlusal devices is fabricated from hard acrylic resin with canine guidance and an anterior plateau.[1,2]

The traditional methods for fabricating these devices have been vacuum thermoforming, sprinkling of acrylic resin,[3] or a combination of both.[4,5] Meanwhile, computer-aided design and computer-aided manufacturing (CAD-CAM) technology allows the subtractive fabrication of occlusal devices by milling them from a polymer blank. Owing to a high conversion rate of double bonds of the industrially manufactured polymethylmethacrylate (PMMA) blanks, the milled objects have a better fit because of the lack of polymerization shrinkage. An advantage of the digital approach is that occlusal devices are manufactured faster and more consistently than manual fabrication. Because

ABSTRACT

Statement of problem. Polymeric material for 3-dimensional printing can be used to fabricate occlusal devices. However, information about fracture resistance and wear is scarce.

Purpose. The purpose of this in vitro study was to investigate the fracture resistance and 2-body wear of 3-dimensional−printed (3DP) (FotoDent splint; Dreve Dentamid GmbH), milled polymethylmethacrylate (CAM) (Temp Basic; Transpa 95H16, Zirkonzahn GmbH), and conventionally fabricated polymethylmethacrylate (CAST) (Castdon; Dreve Dentamid GmbH) occlusal devices.

Material and methods. A total of 96 occlusal devices were prepared according to the 3 different manufacturing techniques 3DP, CAM, and CAST (n=32). For each manufacturing technique, specimens were further divided into initial fracture resistance tests (n=16) and artificial aging in the mastication simulator (50 N, 37°C) with 2-body wear followed by fracture resistance tests (n=16). The fracture resistance was determined using a universal testing machine (1 mm/min). The wear was measured after 20 000 and 120 000 mastication cycles with the replica technique, mapped with a laser scanner, and quantified in R software. Data were analyzed using a 2-way ANOVA followed by a 1-way ANOVA with Scheffé or Games-Howell post hoc tests, repeated measures ANOVA with corrected Greenhouse-Geisser P values, and the Levene, Mann-Whitney, and paired t tests (α=.05).

Results. CAM presented higher initial fracture resistance than 3DP or CAST ($P<.001$). After mastication simulation, CAM followed by 3DP showed higher fracture resistance than CAST ($P<.001$). Mastication simulation decreased the fracture resistance for CAM and CAST ($P<.001$) but not for 3DP ($P=.78$). Three-dimensional−printed occlusal devices showed the highest material volume loss, followed by CAM and the lowest in CAST ($P<.001$).

Conclusions. Three-dimensional−printed occlusal devices showed lower wear resistance and lower fracture resistance than those milled or conventionally fabricated. Therefore, only short-term application in the mouth is recommended. Further developments of occlusal device material for 3-dimensional printing are necessary. (J Prosthet Dent 2019;121:166-72)

[a]Graduate student, Dental Material Unit, Department of Prosthetic Dentistry, Ludwig-Maximilians-University, Munich, Germany.
[b]Research Fellow, Dental Material Unit, Department of Prosthetics, Ludwig-Maximilians University, Munich, Germany.
[c]Senior Statistician, Department of Biostatistics, Epidemiology, Biostatistics and Prevention Institute, University of Zurich, Zurich, Switzerland.
[d]Research Associate, Dental Material Unit, Department of Prosthodontics, Ludwig-Maximilians University, Munich, Germany.
[e]Dental Technician, Dental Material Unit, Department of Prosthetic Dentistry, Ludwig-Maximilians University, Munich, Germany.
[f]Scientific Head, Dental Material Unit, Department of Prosthodontics, Ludwig-Maximilians University, Munich, Germany.

Clinical implications

With current technology, 3-dimensional–printed occlusal devices should be limited to short-term application only. 3-D printed material showed lower wear resistance than milled or injection molded parts.

occlusal devices generally cover a complete arch, the designs do not nest well in a round acrylic resin blank. Hence, a maximum of 2 occlusal devices can be milled from 1 blank, causing considerable waste.

A relatively new approach to avoid the waste of materials is additive manufacturing technology. Although introduced in 1986 by Charles W. Hull,[6] it took some time until additive manufacturing, more commonly referred to as 3-dimensional (3D) printing, was adopted by dentistry.[7,8] Several additive techniques have been successfully established in dentistry, including the fabrication of crowns, fixed partial dentures,[9] and partial denture frameworks[10] by using selective laser melting or sintering.[8] It is not only possible to print metal powders but also to form polymers with fused deposition modeling and polymerize polymeric resins using ultraviolet-light with layering techniques including stereolithography (SLA),[11] PolyJet, and digital light processing (DLP).[12] These approaches allow the fabrication of casts for occlusal device fabrication by vacuum thermoforming.[13] Interim crowns have recently been fabricated using DLP.[14] Adding ultraviolet absorbers to the resin allows the printing of clear objects.[12] This is particularly interesting for the fabrication of occlusal devices, implant drilling guides, and transparent surgery models. Combined with computed tomography or cone beam computed tomography data, precise casts can be manufactured.[15]

In maxillofacial surgery, occlusal devices have been successfully applied to reposition the jaws in orthognathic surgery.[16,17] Printed drilling guides have also been successful in implant dentistry.[18] However, 3D-printed (3DP) occlusal devices for patients with bruxism are used for longer periods and must withstand forces of up to 770 N.[19] Such forces can be developed during tooth grinding and exceed normal mastication forces.[20] Even though printing accurate occlusal devices is feasible,[21] only 1 study has investigated the wear resistance of DLP-printed occlusal devices[22] and another the mechanical properties of SLA-printed occlusal devices.[23] Information about the fracture resistance of 3DP occlusal device materials is still lacking.

Therefore, the purpose of this in vitro study was to evaluate the wear and fracture resistance of 3DP occlusal devices and compare these with those of milled PMMA occlusal devices and manually fabricated occlusal devices. This study assumed that all tested occlusal device materials would show comparable material volume loss and

comparable fracture resistance regardless of the mastication cycles.

MATERIAL AND METHODS

Specimens (N=96, n=32) were prepared by using 3 different manufacturing techniques (Table 1; Fig. 1). For this, a simplified crown-like coping occlusal device was fabricated (Dental Designer; 3Shape A/S) and used as a master standard tessellation language (STL) file to fabricate all the coping devices tested. The 3DP specimens (n=32) were fabricated using a DLP 3D printer at 405 nm. Postprocessing procedure, including rinsing in isopropanol and post-polymerization (HiLite Power; Kulzer GmbH), was performed. Coping occlusal devices were milled from a PMMA blank (CAM) (Temp Basic Transpa; Zirkonzahn GmbH) and from a wax blank (CAST) (Wax ivory, 98H16; Zirkonzahn GmbH) using milling units (M1; Zirkonzahn GmbH; i-mes, Wieland Dental & Technik GmbH & Co KG). For CAST, wax-coping occlusal devices were further embedded in a flask with gypsum for the injection technique. The resin (Castdon; Dreve Dentamid GmbH) was mixed according to the manufacturer's instructions, injected (Palajet; Kulzer GmbH) at 0.5 MPa, further polymerized (at 0.2 MPa, 55°C, and 30 minutes using Palamat elite; Kulzer GmbH), and deflasked. All specimens were polished using 3 polishing steps with decreasing particle size from coarse (95 μm), to medium (50 μm), to fine (5 μm) (9424, 9432, 9433, Kunststoff-Polierer Handstück; Gebr Brasseler GmbH & Co KG). High gloss was achieved using polish (Polishing Brushes; Polirapid) with a goat hair brush and a high-luster buff. The specimens were adhesively cemented[24] (SmartCem2; Dentsply Sirona) onto tapered metal alloy abutments. Before cementation, the abutments were airborne-particle abraded (0.2 MPa, 10 seconds, Renfert basic Quattro IS; Renfert GmbH) with 50-μm alumina powder (Orbis Dental Handelsgesellschaft mbH) and cleaned in distilled water for 10 minutes using an ultrasonic device (L&R Transistor/Ultrasonic T-14; L&R). The coping occlusal devices were then mounted in a cementing device to ensure a specific, centralized load of 7.4 N for 2 minutes. Polymerization was performed from mesial, distal, buccal, lingual, and occlusal surfaces for 40 seconds on each specimen using an light-emitting diode–light polymerization unit (Elipar 2; 3M). Thereafter, the specimens were stored in distilled water at 37°C (Hera cell; Kulzer GmbH) for 24 hours.

For half of the specimens of each manufacturing technique (n=16), the initial fracture resistance was measured using a universal testing machine (Retro Line; Zwick Roell) with a ball of 8-mm diameter at the center of the coping occlusal device. To avoid force peaks, a

Table 1. Technologies, manufacturer's name, lot numbers of tested materials, systems used, and process parameters

Manufacturing Technology	Abbreviation	Material (Manufacturer)	Lot No.	System (Manufacturer)	Process Parameters
DLP printing	3DP	Photosensitive resin (FotoDent splint; Dreve Dentamid GmbH)	JG-1-87-1	DLP 3D printer at 405 nm (Dreve Dentamid GmbH)	Layer thickness: 50 μm
					Resolution: 60 μm
					Wavelength: 405 nm
Milling	CAM	Temp Basic Transpa 95H16 (Zirkonzahn GmbH)	8238, 8104	M1 (Zirkonzahn GmbH)	Five axis
					Spindle speed: maximum of 50 000 U/min
Injection molding	CAST	PMMA (Caston; Dreve Dentamid GmbH)	2016009394 2016009078	Palamat Elite (Kulzer GmbH)	Temperature: 55°C
					Pressure: 0.2 MPa
					Duration: 30 min

DLP, digital light processing; PMMA, polymethylmethacrylate.

Teflon foil with a thickness of 0.1 mm (Angst+Pfister) was placed between the pontic and the loading device. The first investigation of the surface was performed after loading with 800 N (preloading) to determine the initial formation of cracks in the occlusal device material. Every coping was then photographed to document possible cracks or fractures. Subsequently, the coping occlusal devices were loaded till failure of the material appeared, or measurements were automatically stopped when 20% of the maximum force was reached with an overall maximum load of 7500 N.

The remaining coping occlusal devices were artificially aged in a mastication simulator[25] (CS-4; SD Mechatronik GmbH) with enamel antagonists, isothermally at a temperature of 37°C in distilled water, for 120 000 mastication cycles, at a frequency of 1.1 Hz, with 50 N[26-29] vertical load, and 0.7-mm horizontal sliding movement. The enamel antagonists were prepared from human maxillary molars donated by anonymous patients in the Munich area. All teeth were stored in chloramine solution (0.5% Chloramine-T; Sigma-Aldrich Corp) for 7 days and stored in distilled water at 5°C for a maximum of 6 months afterward. The mesiobuccal cusp of each tooth was separated and provided with apical retentions for embedding them in a stainless steel pattern with amalgam (Dispersalloy; Dentsply Sirona). Standardization of the shape was achieved by drilling the cusps in a dome-like contour using a concave drill with grainsize of 40 μm and 8 μm (Gebr. Brasseler GmbH & Co KG).

For the 2-body wear analyses,[30-32] impressions were made (Flexitime Fast and Scan; Kulzer GmbH) at 3 successive points in time (before mastication simulation, after 20 000 mastication cycles, and 120 000 mastication cycles). The impressions were digitized using a laser scanner (LAS-20; SD Mechatronik GmbH). All scans were conducted by a trained operator (A.-M.L.). The scanning parameters were as follows: vertical resolution of 0.8 μm, horizontal resolution of 0.2 μm, and measuring field of 5×5 mm. Thereafter, the fracture resistance of the aged specimens was determined.

Material volume loss was quantified using the R software (R Foundation for Statistical Computing; www.r-project.org) applying the geometric approach.[33] Descriptive statistics, including mean and standard deviation, were computed. The assumption of the approximate normality of measurements was evaluated by the Kolmogorov-Smirnov test. The 2-way ANOVA was applied to determine the interaction between the material and aging-wear condition with respect to the fracture resistance. For each aging-wear condition separately, a 1-way ANOVA with Scheffe post hoc tests was applied to determine differences in fracture resistance between the material groups. The repeated measures ANOVA with corrected Greenhouse-Geisser P values was applied to determine the longitudinal effect of the number of cycles on the material volume loss in different material groups. The Levene test was applied to evaluate the assumption of homoscedasticity of variance in ANOVA setting. The Mann-Whitney test was applied to determine differences in fracture resistance between 2 differing aging conditions. A paired t test was performed to determine the effect of the increasing number of mastication cycles on the material volume loss for each material group separately (α=.05).

RESULTS

Different techniques of manufacturing and mastication simulation significantly influenced the fracture resistance (P=.010; Table 2). Initial fracture resistance was significantly higher in CAM than in CAST and 3DP (P<.001; Table 3). After mastication simulation with 120 000 cycles, the highest fracture resistance values were observed for CAM, then for 3DP, followed by CAST (P<.001). CAM and CAST demonstrated significantly higher initial
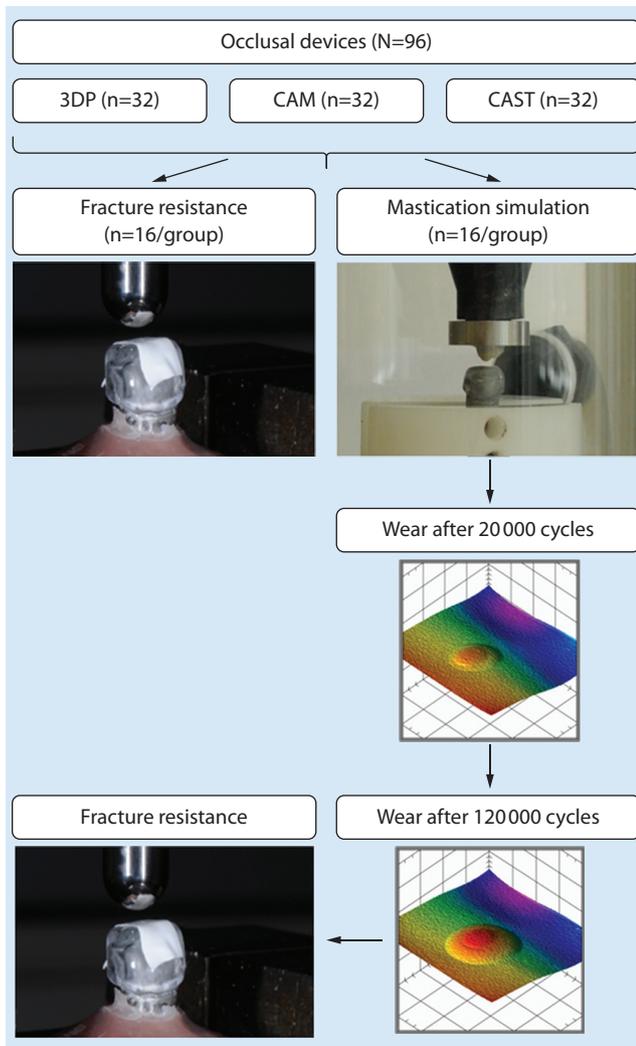
Figure 1. Study procedures.

fracture resistance values than after mastication simulation ($P<.001$), but not 3DP ($P=.78$).

The repeated ANOVA measurements revealed significant differences among the tested materials ($P<.001$; Table 4) and an interaction between the number of mastication cycles and the material volume loss ($P<.001$). Increasing numbers of mastication cycles significantly increased the material volume loss of all tested materials ($P<.001$). The Levene test indicated differences in variance among the tested materials ($P<.001$), making the Games-Howell post hoc test necessary. At both the timepoints (Table 2), 3DP showed the highest material volume loss, and CAST presented the lowest material volume loss ($P<.001$). CAST also presented the lowest increase between 20 000 and 120 000 mastication cycles ($P<.001$).

After mastication simulation, only 3DP showed cracks in every specimen, whereas CAM and CAST presented no cracks in the material. Preloading of initial specimens revealed material cracks in all 3DP specimens, and also in 7 of CAM and 3 of CAST specimens. Preloading of mastication-simulated specimens with 800 N led to material cracks in every specimen of each material, except for 1 CAM specimen.

DISCUSSION

This study investigated the 2-body wear of 3DP, milled, and conventionally fabricated occlusal device materials after 20 000 and 120 000 cycles in a mastication simulator as well as the initial fracture resistance after 120 000 mastication cycles. The overall results demonstrated significant differences between the investigated materials and fabrication techniques. This observation is important in evaluating the suitability of the 3DP material for the fabrication of occlusal devices and for comparing the applications.

The results of high fracture resistance for the milled coping occlusal devices were expected and are explained by the industrial fabrication of the PMMA-based CAD-CAM blanks. The material used for the conventional fabrication is also based on PMMA, but because of the manual process, it is more vulnerable to operator influence, which may lower the conversion rates of double bonds, voids, and inhomogeneity. Another explanation of differences in fracture resistance between conventionally fabricated and milled specimens might be the cementation of the coping occlusal devices because an effect of the cement thickness was found on fracture resistance. An increased cement layer thickness resulted in reduced values of fracture resistance.[24] The intaglio of the conventionally fabricated specimens had to be manually fitted onto the abutments to ensure excellent fit. Thus, additional space for an increased cement layer thickness resulted in decreased fracture resistance. Rosentritt et al[25] recently demonstrated that the type of cement affects fracture resistance of CAD-CAM–fabricated interim molar crowns made of PMMA. However, occlusal devices are not cemented onto the teeth and are only worn at night. Therefore, the cementation of the coping occlusal devices is a limitation of this study but was necessary to prevent the lateral movement in the mastication simulator from shifting the coping occlusal devices. This might have influenced the fracture resistance results.

For the 3DP material, only limited information about the chemical composition is provided by the manufacturer. In comparison with the other investigated materials, a different monomer (urethane dimethacrylate) is the main component. Differences in mechanical properties between materials containing PMMA or urethane dimethacrylate may be responsible for observed differences in fracture resistance and material volume loss. Despite the high material volume loss observed in the

Table 2. Two-way ANOVA results for comparison of fracture load after different manufacturing techniques and mastication simulation

Source	Sum of Squares	df	Mean Squares	F	P
Constant term	588331786	1	588331786	2823.239	<.001
Manufacturing technique (materials)	18329128	2	9164564	43.98	<.001
Mastication simulation	4510724	1	4510724	21.65	<.001
Manufacturing technique (materials)×mastication simulation	2016469	2	1008234	4.84	.010
Error	18755005	90	208389	—	—
Total	631943112	96	—	—	—

R square=.570 (corrected R square=.546).

Table 3. Descriptive statistics mean ±standard deviation for initial and final fracture resistance (N) and for material volume loss (mm^3) after 20 000 and 120 000 mastication cycles for each manufacturing technique

Manufacturing Techniques	Fracture Resistance (Initial)	Fracture Resistance (after 120 000 Mastication Cycles)	Material Volume Loss (after 20 000 Mastication Cycles)	Material Volume Loss (after 120 000 Mastication Cycles)
3DP	2286 ±499	2263 ±450	1.2 ±0.3	2.8 ±1.0
CAM	3398 ±435	2766 ±528	0.7 ±0.1	1.8 ±0.4
CAST	2393 ±451	1748 ±357	0.4 ±0.1	1.2 ±0.4

Table 4. Repeated measured ANOVA results with corrected Greenhouse-Geisser P values for comparison of material volume loss

		Sum of Squares	df	Mean Squares	F	P
Mastication cycles	Greenhouse-Geisser	34.7	1.000	34.7	249	<.001
Mastication cycles×Manufacturing technique (materials)	Greenhouse-Geisser	3.2	2	1.6	11.6	<.001
Error (Mastication cycles)	Greenhouse-Geisser	6.254	45	0.139	—	—

Except for initial fracture resistance and material volume loss after 120 000 cycles in 3DP, assumption of approximate normality not rejected by Kolmogorov-Smirnov test.

3DP material, the fracture resistance was not affected by mastication simulation. This may indicate high reliability, good resistance against aging, and good homogeneity of the material based not only on the chemical composition but also on the processing parameters of 3D printing.

Different printing directions influence the accuracy of DLP-printed objects,[14] and the flexural strength was highest in SLA-printed occlusal devices with vertical printing direction.[23] The specimens used in this study were printed vertically. In addition, decreasing layer thickness enhances the flexural strength of SLA-printed objects.[11] This may also have affected the fracture resistance of DLP-printed objects because both the techniques polymerize acrylic resins on a layer-by-layer basis.[12]

The tested materials exhibited different fracture patterns that are shown in Figure 2. All showed crazing, but the 3DP material and the conventionally fabricated material exhibited a more brittle fracture with less plastic deformation than specimens milled from PMMA-based CAD-CAM blanks. This is associated with varying ductility, which was confirmed in additional internal Martens hardness measurements. Here, the highest indentation modulus was found for the conventionally fabricated coping, followed by 3DP copings.

The outcome for material volume loss differs from that of the study by Huettig et al,[22] who stated that wear resistance was comparable in conventional, subtractive,

and additive manufactured specimens. In contrast with the present study, a load of only 5 N and 5000 cycles was applied, which may have been insufficient loading and too few cycles to detect substantial differences. In general, material volume loss increases with the number of mastication cycles.[30] Thus, a higher number of cycles would probably illustrate differences among the materials more clearly. In this study, wear was measured after 20 000 and 120 000 cycles, which corresponds to a night-time use of the occlusal device for 1 and 6 months, respectively; this was calculated based on a maximum number of 1400 mastication cycles per day.[34] Occlusal devices are usually worn at night when no variations in temperature occur because of the ingestion of cold and hot food or drinks. Therefore, this study was conducted isothermally at 37°C. This is in accordance with previous studies.[31,32] A further testing parameter different from that of Huettig et al[22] was the vertical load of 50 N, which is used as a standard.[26-29] In general, a direct comparison of wear resistance is difficult because test arrangements, parameters, and tested occlusal device materials varied widely among different studies.

To evaluate the material volume loss, a replica technique was developed for this study. This allowed longitudinal wear mapping using a dedicated laser scanner without removing specimens from the mastication simulator during testing and its quantification by the geometric approach using the R software.[33] Other studies have also used 3D laser scanners for volume loss
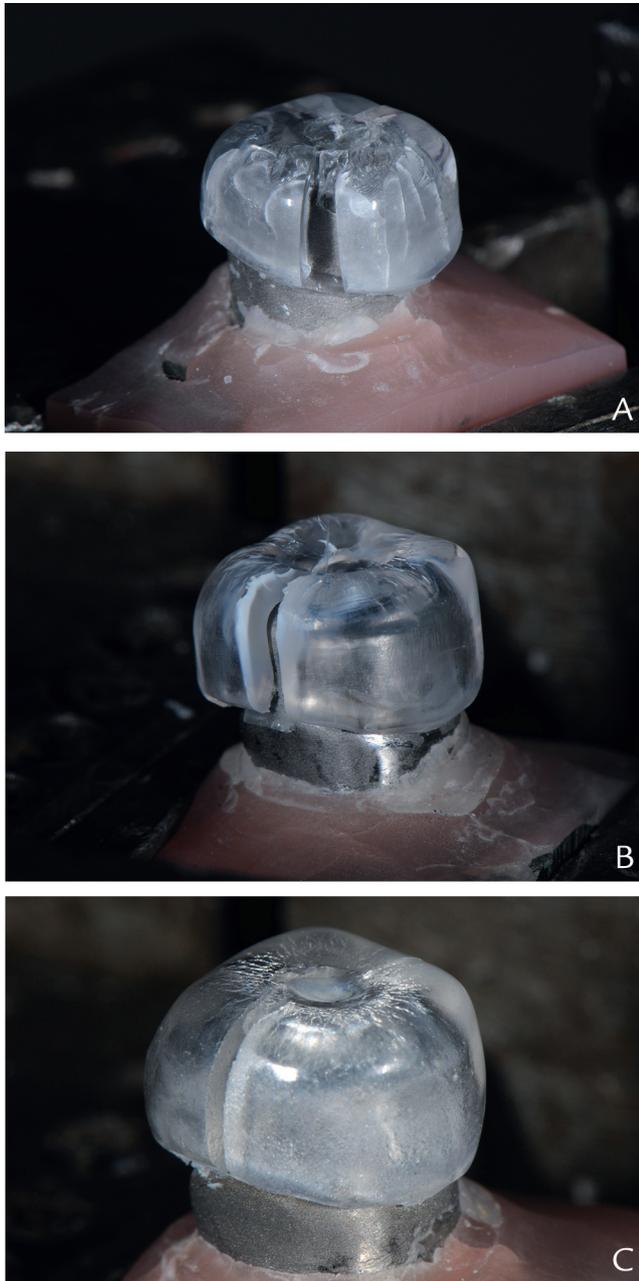
Figure 2. Typical fracture images of occlusal devices after fracture resistance tests: A, Three-dimensional printed; B, Milled; C, Conventionally fabricated.

evaluation[26] because optical methods are found to be more efficient.[30]

Further studies with equal testing arrangements, parameters, and evaluation techniques should be conducted to allow the comparison of research results. In addition, clinical studies should examine the overall wear of printed occlusal devices because clinical occlusal device wear was found to be located unevenly and asymmetrically,[35] something which cannot be addressed within the limitations of an in vitro study.

Even if the 3DP material wore out the most, it showed comparable fracture resistance after an equivalent application time of 6 months, indicating that this material may be used clinically longer than 1 month.

CONCLUSIONS

Based on the fracture resistance and 2-body wear data measured in this study, the following conclusions were drawn:

1. Three-dimensional printed occlusal devices showed lower wear resistance and lower fracture resistance than those milled or conventionally fabricated.
2. Material for 3DP occlusal devices can be used clinically for 1 month. This is in accordance with the period for which the material is currently approved.

REFERENCES

1. Klasser GD, Greene CS. Oral appliances in the management of temporomandibular disorders. Oral Surg Oral Med Oral Pathol Oral Radiol Endod 2009;107:212-23.
2. Dao TT, Lavigne G. Oral splints: the crutches for temporomandibular disorders and bruxism? Crit Rev Oral Biol Med 1998;9:345-61.
3. Kass CA, Tregaskes JN. Occlusal splint fabrication. J Prosthet Dent 1978;40:461-3.
4. Becker CM, Kaiser DA, Lemm RB. A simplified technique for fabrication of night guards. J Prosthet Dent 1974;32:582-9.
5. Bohnenkamp DM. Dimensional stability of occlusal splints. J Prosthet Dent 1996;75:262-8.
6. Hull CW. inventor; Google Patents, assignee. Apparatus for production of three-dimensional objects by stereolithography. 1986.
7. Qingbin L, Leu MC, Schmitt SM. Rapid prototyping in dentistry: technology and application. Int J Adv ManufTechnol 2006;29:317-35.
8. van Noort R. The future of dental devices is digital. Dent Mater 2012;28:3-12.
9. Tara MA, Eschenbach S, Bohlsen F, Kern M. Clinical outcome of metal-ceramic crowns fabricated with laser-sintering technology. J Prosthet Dent 2011;107:62.
10. Williams RJ, Bibb R, Eggbeer D, Collis J. Use of CAD/CAM technology to fabricate a removable partial denture framework. J Prosthet Dent 2006;96:96-9.
11. Chockalingam K, Jawahar N, Chandrasekhar U. Influence of layer thickness on mechanical properties in stereolithography. Rapid Prototyp J 2006;12:106-13.
12. Stansbury JW, Idacavage MJ. 3D printing with polymers: challenges among expanding options and opportunities. Dent Mater 2016;32:54-64.
13. Jiménez M, Romero L, Domínguez Somonte M, Espinosa MM. Rapid prototyping model for the manufacturing by thermoforming of occlusal splints. Rapid Prototyp J 2015;21:56-69.
14. Osman RB, Alharbi N, Wismeijer D. Build angle: does it influence the accuracy of 3D-printed dental restorations using digital light-processing technology? Int J Prosthodont 2017;2:182-8.
15. Rengier F, Mehndiratta A, von Tengg-Kobligk H, Zechmann CM, Unterhinninghofen R, Kauczor H-U, et al. 3D printing based on imaging data: review of medical applications. Int J Comput Assist Radiol Surg 2010;5:335-41.
16. Shaheen E, Sun Y, Jacobs R, Politis C. Three-dimensional printed final occlusal splint for orthognathic surgery: design and validation. Int J Oral Maxillofac Surg 2017;46:67-71.
17. Yanping L, Shilei Z, Xiaojun C, Chengtao W. A novel method in the design and fabrication of dental splints based on 3D simulation and rapid prototyping technology. Int J Adv Manuf Technol 2006;28:919-22.
18. Flügge TV, Nelson K, Schmelzeisen R, Metzger MC. Three-dimensional plotting and printing of an implant drilling guide: simplifying guided implant surgery. J Oral Maxillofac Surg 2013;71:1340-6.
19. Nishigawa K, Bando E, Nakano M. Quantitative study of bite force during sleep associated bruxism. J Oral Rehabil 2001;28:485-91.
20. Waltimo A, Könönen M. A novel bite force recorder and maximal isometric bite force values for healthy young adults. Eur J Oral Sci 1993;101:171-5.
21. Salmi M, Paloheimo K-S, Tuomi J, Ingman T, Mäkitie A. A digital process for additive manufacturing of occlusal splints: a clinical pilot study. J R Soc Interface 2013;10:20130203.

22. Huettig F, Kustermann A, Kuscu E, Geis-Gerstorfer J, Spintzyk S. Polish-ability and wear resistance of splint material for oral appliances produced with conventional, subtractive, and additive manufacturing. J Mech Behav Biomed Mater 2017;75:175-9.
23. Väyrynen VOE, Tanner J, Vallittu PK. The anisotropicity of the flexural properties of an occlusal device material processed by stereolithography. J Prosthet Dent 2016;116:811-7.
24. Gressler May L, Kelly JR, Bottino MA, Hill T. Influence of the resin cement thickness on the fatigue failure loads of CAD/CAM feldspathic crowns. Dent Mater 2015;31:895-900.
25. Rosentritt M, Raab P, Hahnel S, Stöckle M, Preis V. In-vitro performance of CAD/CAM-fabricated implant-supported temporary crowns. Clin Oral Investig 2017;21:2581-7.
26. Kurt H, Erdelt KJ, Cilingir A, Mumcu E, Sülün T, Tuncer N, et al. Two-body wear of occlusal splint materials. J Oral Rehabil 2012;39:584-90.
27. Hahnel S, Behr M, Handel G, Rosentritt M. Two-body wear of artificial acrylic and composite resin teeth in relation to antagonist material. J Prosthet Dent 2009;101:269-78.
28. Lambrechts P, Debels E, Van Landuyt K, Peumans M, Van Meerbeek B. How to simulate wear? Dent Mater 2006;22:693-701.
29. Wimmer T, Huffmann AMS, Eichberger M, Schmidlin PR, Stawarczyk B. Two-body wear rate of PEEK, CAD/CAM resin composite and PMMA: effect of specimen geometries, antagonist materials and test set-up configuration. Dent Mater 2016;32:e127-36.
30. Heintze SD. How to qualify and validate wear simulation devices and methods. Dent Mater 2006;22:712-34.
31. Casey J, Dunn WJ, Wright E. In vitro wear of various orthotic device mate-rials. J Prosthet Dent 2003;90:498-502.
32. Schulte JK, Anderson GC, Sakaguchi RL, DeLong R. Wear resistance of Isosit and polymethyl methacrylate occlusal splint material. Dent Mater 1987;3:82-4.
33. Stöckl C, Hampe R, Stawarczyk B, Haerst M, Roos M. Macro- and microtopographical examination and quantification of CAD-CAM composite resin two- and three- body wear. J Prosthet Dent 2018; accepted.
34. Kelly JR. Ceramics in restorative and prosthetic dentistry. Annu Rev Mater Sci 1997;27:443-68.
35. Korioth TWP, Bohlig KG, Anderson GC. Digital assessment of occlusal wear patterns on occlusal stabilization splints: a pilot study. J Prosthet Dent 1998;80:209-13.

Corresponding author:
Dr Bogna Stawarczyk
Goethestraße 70, 80336 Munich
GERMANY
Email: bogna.stawarczyk@med.uni-muenchen.de