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Influence of CAD/CAM Fabrication and Sintering Procedures on the Fracture Load of Full-Contour Monolithic Zirconia Crowns as a Function of Material Thickness

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Clinical Relevance

Chairside sintering might be a suitable procedure to produce zirconia restorations with clinically adequate fracture loading values for specific restoration thicknesses.

Summary

Objective: The purpose of this *in vitro* study was to analyze the effect of computer-aided design/computer-aided manufacturing (CAD/CAM) fabrication and sintering procedures on the fracture load of monolithic zirconia crowns with different material thicknesses adhesively seated to methacrylate dies fabricated with stereolithography technology.

Method: Monolithic zirconia crowns were fabricated from inCoris TZI C material with a chairside CAD/CAM system (CEREC MCXL) comprising three material thicknesses (0.5/1.0/1.5 mm, n=8 each). Two CAD/CAM fabrication procedures (milling, MI; grinding, GR), two chairside sintering procedures (superspeed, SS; speedfire sintering, SF), and one labside sintering procedure (classic, CL) were evaluated. In total, 144 crowns were fabricated. Restorations were adhesively seated to methacrylate dies fabricated with SLA technology. Thermomechanical cycling (TCML) was performed before fracture testing. Loading forces until fracture were registered and statistically analyzed with one-way analysis of variance (ANOVA), *post hoc* Scheffé test, and three-way ANOVA (α=0.05).

Results: Test groups showed statistically significant differences ($p<0.05$). The highest mean value was found for 1.5-mm crowns of

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group GR_SF with 3678.6 ± 363.9 N. The lowest mean value was found for group 0.5-mm crowns of group MI_SF with 382.4 ± 30.7 N. There was a significant three-way interaction effect between thickness, sintering, and processing [$F(4,126)=9.542$; $p<0.001$; three-way ANOVA, significance level $\alpha=0.05$].

Conclusions: CAD/CAM fabrication and sintering procedures influence the maximum loading force of monolithic zirconia crowns with different material thicknesses. A material thickness of 0.5 mm should be considered as a critical thickness for monolithic zirconia crown restorations.

Table 1: *Overview Test Groups*

Group	Fabrication Procedure	Sintering Procedure	Sintering Device
MI_CL	Milling	Classic	inFire HTC speed
GR_CL	Grinding		
MI_SS	Milling	Superspeed	
GR_SS	Grinding		
MI_SF	Milling	Speedfire	CEREC SpeedFire
GR_SF	Grinding		

Monolithic zirconia material inCoris TZI C was used for all groups; for each group three different subgroups with material thicknesses 0.5/1.0/1.5 mm were established with each (n=8). A total of n = 144 crowns were fabricated. There were two different CAM fabrication processes (MI = milling, GR = grinding) using a CEREC MCXL CAM device and three different sintering processes (CL = classic, SS = superspeed, SF = speedfire).

INTRODUCTION

Zirconia ceramic materials have been described in literature both for single restoration and fixed dental prostheses (FDPs) indications.[1] Y-TZP (yttria-stabilized tetragonal zirconia polycrystal) is the most widely used type of zirconia material.[2] Marginal adaptation, internal fitting, and clinical long-term survival of zirconia restorations have been demonstrated to be clinically acceptable compared with conventional ceramic materials.[3,4] Zirconia materials have advantageous material characteristics such as a superior fracture resistance and a higher fracture toughness.[5] The extent of tooth preparation might be thus reduced for zirconia restorations.[6,7] Among the reported shortcomings for zirconia are a high opacity, a reduction of fracture strength as a result of low-temperature aging degradation, and the possibility of chipping of veneering ceramic.[8–12] Zirconia materials show superior wear behavior and lower antagonist wear compared with glass ceramics after grinding treatments.[13] Until now, few clinical studies are available investigating the clinical long-term success of monolithic zirconia restorations, and a recommended minimum thickness for monolithic zirconia restorations has not been reported.[14–16]

Zirconia restorations are normally fabricated in a presintered green body state with a geometric magnification of up to 20% to 30% using subtractive computer-aided design/computer-aided manufacturing (CAD/CAM) technology.[1] Subtractive fabrication can be performed either with milling technology using carbide burs or with grinding technology using diamond-coated instruments. Both fabrication methods can be executed in dry and wet surroundings. Most zirconia restorations today are produced using dry-milling CAM fabrication procedures. Few studies are available investigating a possible effect of the respective CAM fabrication method on the material characteristics of zirconia.[17,18] Zirconia restorations obtain their final geometric form during a subsequent high temperature sintering process using special ceramic furnacing devices. Various sintering procedures are available for zirconia restorations both varying the sintering time and the sintering temperature. Few studies have investigated a possible effect of sintering procedures on the material characteristics of zirconia.[19–22]

The aim of this study was to analyze the effect of different fabrication and sintering procedures on the fracture load of full-contour monolithic zirconia crowns with different material thicknesses *in vitro*. The null hypothesis was that different sintering and fabrication procedures do not influence the maximum fracture load of monolithic zirconia crowns with different material thicknesses.

METHODS AND MATERIALS

In this study, fatigue loading and subsequent fracture loading of monolithic zirconia crowns adhesively seated to abutment dies fabricated with stereolithography (SLA) technology was performed. Three different crown material thicknesses (0.5/1.0/1.5 mm) were investigated for each specific test group. Monolithic partially stabilized zirconia material (Y-TZP) inCoris TZI C (Dentsply Sirona, York, USA) was used for all test groups. Test groups comprised two different CAM fabrication procedures and three different sintering procedures. For each test group, eight crowns were fabricated. In total, 144 crowns were fabricated. Specification of test groups investigated in this study is shown in Table 1.

All crowns were adhesively seated to abutment dies fabricated with SLA technology. Design of abutment dies was performed in accordance with preparation guidelines for full-ceramic restorations

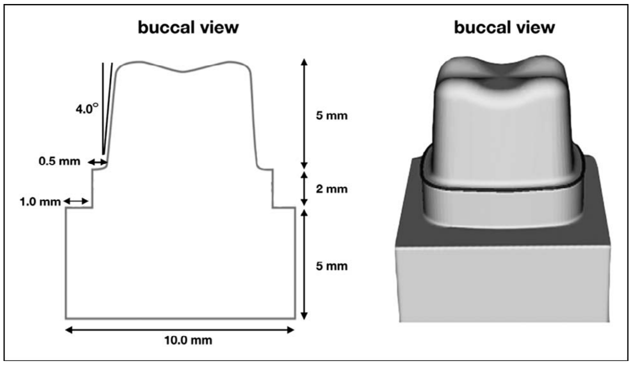

Figure 1. *Design of abutment die for 0.5 mm crown; left: cross section with indication of metric values; right: digitized abutment die with indication of preparation margin highlighted in blue.*

with CAD Software Pro Engineer Wildfire 4.0 (PTC, Boston, USA). Three different die geometries were designed in respect of full-contour crown material thicknesses 0.5/1.0/1.5 mm. The specification of the abutment design was that the restoration's outer contour could be designed identically for each material thickness. SLA printer Viper Si2 (3D Systems, Rock Hill, USA) was used for abutment die fabrication. z-axis solution was 50 μm for the abutment die. A methacrylate material was used as abutment material (E-Modulus, 2.5 GPa; fractural strength, 110-130 MPa; Shore hardness, 80-84 Shore D). The specifications of the designed abutment die for a 0.5 mm crown is shown in Figure 1.

The design of full-contour crowns was performed using CAD software CEREC v.4.0 (Dentsply Sirona) after initially scanning the abutment die for 0.5-mm material thickness with the intraoral scanning device CEREC Bluecam (Dentsply Sirona). Using the software tool "biocopy," the design of the 0.5-mm restoration could be transferred with an identical outer contour but an individual adaptation of the inner contour to the abutment dies for 1.0- and 1.5-mm restorations. Crown material thicknesses were controlled in all cases by applying CAD software tools "cursor details" and "show minimum thickness" so that an identical design could be ensured for the respective test groups. Restoration parameters were identical for all crowns with a spacer parameter setting of 80 μm. Different material thicknesses of 0.5/1.0/1.5 mm always comprised the entire crown restoration.

All crowns were fabricated with a chairside CAD/CAM system using a 3+1 axis milling unit CEREC MCXL (Dentsply Sirona). There were two different CAM fabrication procedures: group MI (milling) and group GR (grinding). The CEREC MCXL milling machine was equipped with different carbide burs for milling (Shaper 25/RZ, Finisher 10) and diamond-coated burs for grinding (StepBur 20, Cylinder Pointed Bur 20). Milling and grinding mode was set to "standard." Milling and grinding instruments were renewed after the fabrication of one test group (ie, after eight restorations to exclude a possible damaging effect of used instruments to the restoration while fabrication).

Monolithic zirconia crowns were forwarded to different sintering procedures after fabrication. There were three different sintering procedures: group CL (classic sintering with inFire HTC speed; Dentsply Sirona), group SS (superspeed sintering with inFire HTC speed; Dentsply Sirona), and group SF (speedfire sintering with CEREC SpeedFire; Dentsply Sirona). All sintering procedure parameters were strictly according to manufacturer's recommendations: CL, sintering temperature 1510°C, dwell time 120 minutes, total time eight hours, start from room temperature; SS, sintering temperature 1580°C, dwell time 10 minutes, total time 10 minutes, start at preheated furnace; SF, sintering temperature 1580°C, dwell time two minutes, total time 13.34 minutes, start from room temperature (example for inCoris TZI C, shade A3). No post-processing was performed to the crowns after the sintering process.

Crowns were cleaned ultrasonically for three minutes and degreased with ethanol after sintering. No phosphoric acid etching cleaning of restorations was performed. SLA-fabricated abutment dies and crowns were first sandblasted (Cojet, 3M, Saint Paul, USA) (diameter $\leq$ 50 μm, 200 kPa) and then cleaned with ultrasonic using ethanol for three minutes. Abutment dies were silanized using Monobond Plus (Ivoclar Vivadent AG, Schaan, Lichtenstein) for 60 seconds and then dried with oil-free air. Crowns were then adhesively seated to SLA fabricated abutment dies following a standardized adhesive protocol using a dual cure luting resin cement (PANAVIA F 2.0, Kuraray Noritake, Tokyo, Japan). An oxygen layer inhibitor material (Oxyguard, Kuraray Noritake) was applied around the cervical margin of the crowns during polymerization. A Satelec MiniLED polymerization lamp (KaVo Dental, Biberach an der Riss, Germany) was used for polymerization of the luting resin cement with 1600 mW/cm^2 from the occlusal, mesial, distal, buccal, and lingual aspects for 60 seconds each. All crowns were then forwarded to fatigue loading followed by fracture loading in a universal testing machine.

Restorations were embedded using a methacrylate material (Paladur; Kulzer, Hanau, Germany) in

Table 2: *Results for Maximum Loading Force of Monolithic Zirconia Crowns With Material Thicknesses 0.5/1.0/1.5 mm (n=8)*

Group	Loading Force				95% Confidence Interval		
	Thickness	Mean	SD	Minimum	Maximum	Lower	Upper
MI_CL	0.5	601.4	100.1	402.0	729.6	517.7	685.1
	1.0	1957.3	190.4	1673.8	2243.8	1798.1	2116.5
	1.5	3149.1	423.9	2738.2	4077.0	2794.7	3503.5
GR_CL	0.5	593.1	79.6	514.4	703.1	526.5	659.6
	1.0	2366.2	229.8	2121.0	2830.1	2174.1	2558.3
	1.5	3119.6	456.0	2643.4	4065.2	2738.4	3500.9
MI_SS	0.5	649.9	112.4	532.0	790.8	555.9	743.9
	1.0	1800.0	127.9	1669.0	2032.9	1693.1	1907.0
	1.5	2793.3	567.8	2075.1	3373.6	2318.6	3268.0
GR_SS	0.5	462.3	69.1	366.0	594.1	404.5	520.0
	1.0	1752.5	401.9	1426.1	2476.7	1416.4	2088.5
	1.5	2455.7	85.5	2250.8	2499.8	2384.2	2527.2
MI_SF	0.5	382.4	30.7	350.1	423.6	356.8	408.1
	1.0	1943.4	268.0	1656.5	2282.1	1719.3	2167.5
	1.5	2841.3	188.8	2603.0	3114.6	2683.4	2999.1
GR_SF	0.5	395.8	75.2	314.1	514.2	332.9	458.7
	1.0	1755.2	156.8	1555.7	1979.3	1624.1	1886.4
	1.5	3678.6	363.9	3206.7	4392.6	3374.4	3982.8

Two different CAM fabrication processes (MI = milling, GR = grinding) and three different sintering processes (CL = classic, SS = superspeed, SF = speedfire) were used.

specially designed test blocks ensuring an identical central fixation of the crowns. Thermomechanical loading (TCML) of full-contour zirconia crowns (1.2 million cycles, 1.7 Hz, invariable occlusal load 49 ± 0.7 N, thermal cycling 5-55°, dwell time 120 seconds, 12,000 cycles, water change time 10 seconds, human natural molar cusp antagonist, load to central fissure) was performed.[23] Crowns were investigated for cracks using a stereomicroscope with 14× magnification and transmitted light (Wild Leitz/M1B, Walter Products, Windsor, Canada) after fatigue loading. Fracture loading was performed using Allround Line z010 testing machine (Zwick Roell AG, Ulm, Germany) (crosshead speed, 1 mm/min; ball diameter, 5 mm). All crowns were loaded until fracture. Maximum loading forces were automatically recorded by the software. Fracture data values obtained were statistically evaluated using one-way analysis of variance (ANOVA), and *post hoc* Scheffé test, and three-way ANOVA (α=0.05) (IBM SPSS Statistics v25.0; IBM, Armonk, USA).

For each sintering group (CL, SS, and SF) a 2-mm-thickness disc (dimensions 2 × 2 mm) was prepared with a low-speed diamond saw (Isomet; Buehler, Lake Bluff, USA) and sintered using the respective sintering parameters. Sintered discs were wolfram sputtered (CCU-010, Safematic GmbH, Bad Ragaz, Switzerland) and forwarded to Scanning Electron Microscopy (SEM; Zeiss Supra V50, Carl Zeiss, Oberkochen, Germany) for qualitative microstructural imaging of the respective material structure.

RESULTS

Results for fracture loading forces of full-contour monolithic zirconia crowns are shown in Table 2. Data for fracture loading were normally distributed with homogeneity of variances (Shapiro-Wilk test and Levene test). Maximum loading forces statistically significantly varied among the different test groups (one-way ANOVA and *post hoc* Scheffé test, $p<0.05$). For 0.5- and 1.0-mm-thickness restorations, both the fabrication procedure (MI and GR) and the sintering procedure (CL, SS, SF) showed no statistically significant effect ($p>0.05$). However, for 1.5-mm-thickness restorations, statistically significant differences were found between test groups with different fabrication and sintering procedures ($p<0.05$). There was a significant three-way interaction effect between thickness, sintering, and processing [$F(4,126)=9.542$; $p<0.001$; three-way ANOVA, significance level α=0.05]. The highest mean value was found for 1.5-mm crowns of group GR_SF with 3678.6 ± 363.9 N. The lowest mean value was found for group 0.5-mm crowns of group MI_SF with 382.4 ± 30.7 N. Based on the results found in this study, the null hypothesis that sintering and

Table 3: *Results for Fracture Loading Force of Monolithic Zirconia Crowns With Material Thicknesses 0.5/1.0/1.5 mm*[a]

Crown Thickness	Sintering and Fabrication Procedures					
	CL		SS		SF	
	GR	MI	GR	MI	GR	MI
0.5	593.1 ± 79.6 A	601.4 ± 100.1 A	462.3 ± 69.1 A	649.9 ± 112.4 A	395.8 ± 75.2 A	382.4 ± 30.7 A
1.0	2366.2 ± 229.8 BC	1957.3 ± 190.4 B	1752.5 ± 401.9 B	1800.0 ± 127.9 B	1755.2 ± 156.8 B	1943.4 ± 268.0 B
1.5	3119.6 ± 456.0 DE	3149.1 ± 423.9 DE	2455.7 ± 85.5 BCD	2793.3 ± 567.8 CD	3678.6 ± 363.9 E	2841.3 ± 188.8 CD

[a] *All values are indicated as mean ± SD in [N]. Statistical analysis with one-way ANOVA and post hoc Scheffé test (significance level α=0.05) is provided. Values with the same letters are not statistically significant different (*$p>0.05$*). There is a significant three-way interaction effect between thickness, sintering, and processing [*$F(4,126)=9.542$*;* $p<0.001$*; three-way ANOVA, significance level a=0.05). Two different CAM fabrication processes (MI = milling, GR = grinding) and three different sintering processes (CL = classic, SS = superspeed, SF = speedfire) were used.*

fabrication procedures do not influence the maximum fracture load of monolithic zirconia crowns with different material thicknesses was rejected. An overview of the statistical results for fracture load values is shown in Table 3.

DISCUSSION

In this study, the fracture load of full-contour monolithic zirconia crowns fabricated with two different fabrication methods, MI and GR, and three different sintering methods, CL, SS, and SF, was investigated as a function of three different material thicknesses (0.5/1.0/1.5 mm). InCoris TZI C zirconia material (Dentsply Sirona) was used for all test groups. Full-contour monolithic zirconia crowns were adhesively luted to methacrylate dies fabricated with SLA technology. Fracture loading was performed subsequent to fatigue loading. Maximum loading forces were statistically analyzed. Statistically significant differences were found for maximum fracture loading values between the different test groups (one-way ANOVA, *post hoc* Scheffé test, $p<0.05$). Several aspects of this study need to be discussed.

In this study, abutment dies with a relatively low E-Modulus of 2.5 GPa were used. In the literature, there is unanimity that the stiffer the abutment die material, the higher the fracture load values for restorations might be. The E-Modulus of human dentin is reported to be between 7 and 13 GPa.[24] It is important to emphasize that the purpose of this study setup was to illustrate a worst-case scenario for the respective restoration material. Therefore, it would seem reasonable that if dentin abutment dies with a higher E-Modulus would have been used in this study, fracture load values for monolithic zirconia crowns might have been higher.

In this study, TCML was performed before load-to-fracture testing. TCML is an important test setup parameter when evaluating zirconia materials. In the literature, the fracture strength of zirconia restorations has been demonstrated to be reduced by cyclic loading procedures.[25] Additionally, low-temperature aging degradation has been reported to reduce the fracture strength of ziroconia.[10] It might thus be concluded that initial fracture load values for monolithic zirconia restorations would have been higher than the fracture load values found in this study.

In the literature, a recommended minimum thickness for monolithic zirconia restorations has not been reported.[16] Several studies are available, indicating that thicknesses of zirconia restorations can be reduced compared with conventional ceramics.[6,7,26] For the zirconia material inCoris TZI C used in this study, the manufacturer's recommendation for minimum material thickness is available describing wall thicknesses of at least 1.0 mm at the lowest point of the main fissure and the cusp and a circular ceramic thickness of 0.8 to 1.00 mm, with a tapering crown edge of 0.5 mm. Sorrentino and others[6] demonstrated that zirconia crown restorations with 0.5-mm thickness configuration showed sufficient fracture resistance. In this study, highest mean fractural loading force values for the 0.5-mm-thickness group was found for group MI_SS with 649.9 ± 112.4 N. Lowest mean fractural loading force values for the 0.5-mm-thickness group was found for group MI_SF 382.4 ± 30.7 N. Assuming a maximum chewing force of 600 N for a patient, results of this study suggest that a minimum thickness of 1.0 mm might be suitable for clinical application of full-contour monolithic zirconia crowns. However, it is important to state that *in vitro* findings cannot generally be transferred to *in vivo* conditions because of a more complex parameter setting. Results of this study demonstrate that fracture loading values are positively associated with restoration thicknesses. The highest mean fractural loading force values for the 1.5-mm test

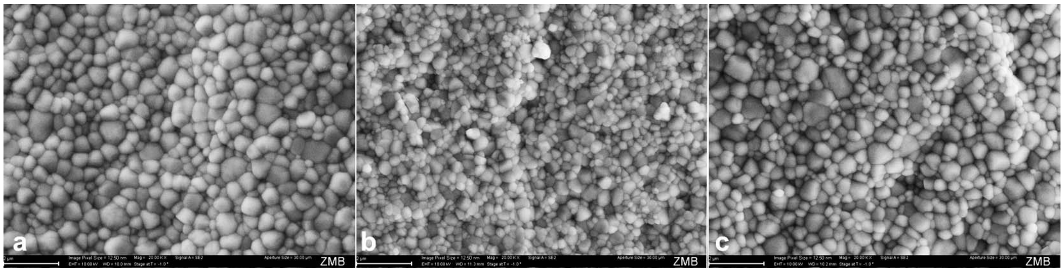

Figure 2. *Qualitative SEM images for specimens sintered with different sintering modes: (A) classic sintering, group CL; (B) speedfire sintering, group SF; (C) superspeed sintering, group SS. SEM images with Zeiss Supra V50 SEM, acceleration speed 10 kV, magnification 20,000×, specimens with wolfram sputtering; notice different grain size and contrast ratio for different sintering protocols.*

group was found for group GR_SF with 3678.6 ± 363.9 N.

In the literature, contradictory results are available for the evaluation of the influence of the fabrication method on material characteristics of zirconia. Ozer and others[16] found that grinding fabrication did not affect the flexural strength of monolithic zirconia. Canneto and others[18] found that grinding fabrication can result in chip damage, with critical flaws subjected to tension and resulting in strength losses. Fraga and others[27] demonstrated that machining negatively affected the flexural strength of zirconia CAD/CAM ceramic up to 40% of a surface-optimized state. Al-Amleh and others[28] reported that zirconia material is sensitive to manufacturing processes that might lead to framework fractures, but the authors do not provide any recommendations about which processing method is least harmful. In the present study, no statistically different results were found between different fabrication procedures for 0.5- and 1.0-mm-thickness groups ($p>0.05$). For 1.5-mm thickness, statistically significant differences were only found within the sintering group SF when comparing milling and grinding procedures ($p<0.05$). Nevertheless, fracture loading values for the 1.5-mm-thickness restorations were within the range of clinical acceptability for both groups. Based on the limitations of this study, results for fracture loading forces thus suggest that both fabrication procedures produce clinically acceptable results for monolithic full-contour zirconia restorations.

In this study, three different sintering procedures were evaluated for the fracture load of monolithic zirconia restorations. For each group sintering temperature, dwell time and total oven time were different. In the literature, several studies are available evaluating the possible effect of sintering procedures on zirconia's material characteristics.[19,20,22,29–31] Kim and others[19] investigated the effects of sintering conditions on grain size and found that grain size increased with prolonged sintering times. Results are thus in good accordance with results found in this study. Qualitative SEM image analysis of specimens using sintering protocols CL, SS, and SF revealed that grain size was highest for group CL and smallest for group SF (Figure 2). SEM image analysis revealed that the distribution of different grain sizes is different for the three sintering protocols. The best homogeneity for the matrix composition was found for group CL.

In the literature, contradictory results are published for the evaluation of different sintering procedures (ie, different grain sizes and matrix compositions of zirconia on the fracture loading force of zirconia restorations). Ebeid and others[20] reported that biaxial flexural strength is not affected by changes of sintering parameters. Sen and others[31] found that increased sintering temperature resulted in increased translucency with minimal impact on biaxial flexural strength. Inokoshi and others[29] reported that higher sintering temperatures and elongated dwell times increased the grain size. The authors concluded that the increase in grain size might favor low temperature degradation of zirconia.[29] Nakamura and others[30] described that mechanical strength of zirconia decreased with increasing grain size. Ersoy and others[22] used the identical zirconia material inCoris TZI C that was used in this study and demonstrated that a combination of high sintering temperatures and short sintering times increased the flexural strength of zirconia. In this study, no statistically significant differences were found within groups 0.5 and 1.0 mm for different sintering protocols ($p>0.05$). Statistically significant differences between different sintering protocols were only found between test groups with 1.5 mm thickness ($p<0.05$). Fracture loading values for 1.5-mm-thickness restorations were within a clinically acceptable range for all three groups. Based on the limitations of this study, the results suggest that sintering procedures CL, SS, and SF produce clinically acceptable results for monolithic full-contour zirconia restorations.

CONCLUSIONS

CAD/CAM fabrication and sintering procedures influence the maximum loading force of monolithic zirconia crowns with different material thicknesses. Chairside sintering might be a suitable procedure to produce zirconia restorations with clinically adequate fracture loading values for specific restoration thicknesses. A material thickness of 0.5 mm should be considered as a critical thickness for monolithic zirconia crown restorations.

Conflict of Interest

The authors of this manuscript certify that they have no proprietary, financial, or other personal interest of any nature or kind in any product, service, and/or company that is presented in this article.

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