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Accuracy of additively manufactured zirconia four-unit fixed dental prostheses fabricated by stereolithography, digital light processing and material jetting compared with subtractive manufacturing

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Declaration of interest:

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submitted version

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Abstract

Objective: To evaluate the manufacturing accuracy of zirconia four-unit fixed dental prostheses (FDPs) fabricated by three different additive manufacturing technologies compared with subtractive manufacturing.

Methods: A total of 79 zirconia FDPs were produced by three different manufacturing technologies, representing additive (one stereolithography [aSLA] and one material jetting [aMJ] device, two digital light processing [aDLP1/aDLP2] devices) and subtractive manufacturing (two devices [s1/s2]), the latter serving as references. After printing, additively manufactured FDPs were debound and finally sintered. Subsequently, samples were circumferentially digitized and acquired surface areas were split in three Regions Of Interest (ROIs: inner/outer shell, margin). Design and acquired data were compared for accuracy using an inspection software. Statistical evaluation was performed using the root mean square error (RMSE) and nonparametric Kruskal-Wallis method with post hoc Wilcoxon-Mann-Whitney U tests. Bonferroni correction was applied in case of multiple testing.

Results: Regardless the ROI, significant differences were observed between manufacturing technologies ($P < 0.001$). Subtractive manufacturing was the most accurate with no significant difference regarding the material/device (s1/s2, $P > 0.054$). Likewise, no statistical difference regarding accuracy was found when comparing s2 with aMJ and aSLA in most ROIs ($P > 0.085$). In general, mean surface deviation was $< 50 \mu\text{m}$ for s1/s2 and aMJ and $< 100 \mu\text{m}$ for aSLA and aDLP2. aDLP1 showed surface deviations $> 100 \mu\text{m}$ and was the least accurate compared to the other additive/subtractive technologies.

Significance: Additive manufacturing represents a promising set of technologies for the manufacturing of zirconia FDPs. Methodological impact on accuracy within and in between different additive technologies needs to be further investigated.

1. Introduction

In recent years, Additive Manufacturing (AM) has become an integral part of various dental applications. In dental laboratories, models, trays and splints are produced from polymers using various AM technologies [1]. Nowadays, metals, e.g. to be used as frameworks are also processed by means of AM [2]. However, there is an increasing demand for metal-free

restorations, resulting in the need for high-strength ceramic materials like oxide ceramics. AM of such ceramics is not yet established in the dental field. In the traditional workflow, various ceramics are available with zirconia being one of the most preferred material [3, 4]. Yttria-stabilized Tetragonal Zirconia Polycrystals containing 3 mol% of Y_2O_3 (often referred as 3Y-TZP) are the most widely used among different zirconia ceramics. They exhibit an excellent combination of fracture toughness ($\sim 5 \text{ MPa}\cdot\text{m}^{1/2}$) and flexural strength ($\sim 1000 \text{ MPa}$) [5]. Stabilizers like yttria are used to retain the tetragonal crystal phase of the zirconia in a metastable state at room temperature. Under the application of mechanical stresses, the tetragonal phase can be transformed to the monoclinic phase, accompanied by a $\sim 4\%$ volume increase which compresses the crack flanks and prevents crack propagation [6].

Nowadays dental zirconia is produced using subtractive manufacturing in a CAD/CAM (computer-aided design and -manufacturing) workflow. Isostatic pressed and pre-sintered blanks with different shades and different flexural strengths are available on the market. Computerized Numerical Controlled (CNC) grinding or milling machines process the restoration from the full material blank, producing a large amount of waste. In contrast, a strong reduction of waste material can be achieved by AM technologies with material only used to build up the part. AM, as a tool-free production technology, also avoid the problem of wear of the tools used in the CNC machine. Some AM technologies, such as Digital Light Processing (DLP), allow for parallel rather than serial manufacturing, which can lead to significant productivity increase. Finally, all these factors can lead to cost and material's waste reduction in the manufacturing of dental prostheses.

In AM of ceramics, different technologies are considered to have the potential to produce monolithic ceramic dental prostheses. Commercially available technologies include Vat polymerization (Stereolithography/SLA and Digital Light Processing/DLP) using light-curable ceramic resin or suspensions as raw materials and Material Jetting (MJ) using ceramic inks [7]. Vat polymerization is the most widely used AM technology for the production of ceramic parts. These processes are based on the local, layerwise photoinitiated polymerization of resins mixed with ceramic particles. The printed part consists of a photopolymerized polymer matrix with embedded ceramic particles. To obtain the final ceramic part, the polymer matrix is burned out in a post-processing step (debinding) and afterwards the part is sintered to obtain a dense monolithic ceramic [8]. Depending on the method, the resin is crosslinked by a laser (SLA) or a digital light projector (DLP). In SLA technologies, ceramic resins or suspensions are treated by a laser beam that traces photopolymerizable areas, layer by layer, following a sliced 3D-model (STL file). The height of a layer typically varies from 10 to 120 μm . After one layer cures,

the building platform is lowered or raised in the z-direction (depending on whether the machine uses a “bottom-up” or “top-down” approach) and this process is repeated as many times as necessary. In DLP, since a projector can expose the entire layer at once, the time required for the illumination step is reduced. In the commonly used setup, the projector illuminates the suspension through a window from the bottom. The growing part hangs onto the building platform and for every layer, it is immersed into the suspension (the fabrication of the part takes place “bottom-up”). The resin between part and window is cured and transformed into a solid layer. For the next layer, the part needs to be separated from the window to allow the resin to refill the gap and afterwards these steps are iterated. Material Jetting (MJ) uses inkjet technology, where the material is jetted through a printhead. In polymer material jetting, photopolymers are used, which also allows multimaterial printing. For AM of ceramics, this technology is used by changing the feedstock to an ink that contains ceramic particles. Thousands of nozzles simultaneously jet the building and support material containing inks onto the building platform, generating a part layer by layer. After the support removal, which can be a water-soluble material, the part is debound and sintered to obtain a the ceramic object [9].

When complex geometries are produced by AM technologies, support structures are a vital component to any part. The support structures to build up an organic-ceramic device such as a four-unit bridge vary for different AM methods due to the different arrangement of printed-parts at the building platform. Support structures on one hand have to provide sufficient retention of the growing part on the building platform and on the other hand need to withstand the peel force/separation force for hanging SLA or DLP systems, in the so-called "bottom-up" arrangement. In processes where the resin is exposed from above ("top-down"), as the 3DCeram system applied in the present study, there is no peel force/separation force. In these cases, support structures are only necessary for overhanging structures. The components can therefore be arranged on the build platform in such a manner that as few support structures as necessary are required. This avoids damage at the surface and reduces post-processing costs. Nevertheless, in vat polymerization, the support structures are made, due to the process, of the identical material as the desired part. Material Jetting, on the other hand, allows the use of different materials and thus support structures can be printed with a material that can be removed with less effort [10], for example a material soluble in water or other solvents.

Besides reliability of the parts and a competitive manufacturing speed, accuracy is the main challenge for the application of AM fabricated ceramic in dentistry. The accuracy of the manufactured parts is of significant importance for the clinical application. Inaccurate marginal fit can be responsible for plaque retention, micro-leakage and cement breakdown [11, 12]. Poor

internal accuracy can increase the thickness of the cement and thus influence the mechanical stability of zirconia-based restorations [13, 14].

Therefore, the aim of the present study was to evaluate the current status of the accuracy of additively compared to subtractive manufactured 3Y-TZP FDPs. For this purpose, FDPs produced with different commercially available additive manufacturing technologies i.e. stereolithography, digital light processing, and material jetting were compared with the reference standard of subtractively produced FDPs using three-dimensional target/actual comparison.

2. Materials and methods

2.1. Sample preparation

The AM printing process started with a 3D model of a FDP that was converted to the standard triangulation or tessellation language (STL format) before being sliced into layers according to each technology requirements. 79 FDPs were produced by different suppliers as follow: Two groups of samples were subtractively manufactured, named s1 (GC, n=16, Zirconia ST, GC, Haasrode, Belgium) and s2 (n=16, priti multidisc ZrO₂ “translucent” Pritidenta, Leinfelden-Echterdingen, Germany); Four groups of samples were produced by three different AM technologies i.e. Stereolithography, named aSLA (n=11, Ceramaker900, 3DCeram, Limoges, France); material jetting, named aMJ (n=16, XJET, Rehovot, Israel) and DLP, named aDLP1 (n=16, 405 nm Prototype DLP-Printer, University of Birmingham, UK) [15] and aDLP2 (n=4, CeraFab System Medical, Lithoz, Vienna, Austria).

Milling of FDPs s1 took 30 min and subsequently the FDPs s1 and s2 were sintered at 1450°C for 6.5h and 2h, respectively.


FDPs of aSLA were printed in 25 µm layers using a commercial 3Y-TZP paste (3D Mix ZrO₂ Zr-P03 grade, 3DCeram) with the use of support structures around the margin and the bottom surface of the pontics that were removed after printing. The pre-set parameters provided by 3DCeram (scraping, hatching and laser power parameters) were employed. The excess, uncured paste was removed using a commercial liquid product (CeraClean, 3DCeram) projected under air pressure. After drying, FDPs were slowly debound in air up to 1150°C and sintered at 1450°C for 2h, following 3DCeram recommendations. No additional post-processing was performed.

The aMJ FDPs were printed using a commercial 3Y-TZP ink (C800 zirconia model dispersion grade 7250001, XJET) using a layer thickness of 10.5 μm . The surrounding support was removed by cleaning the printed bridges in distilled water with a brush. Sintering was performed in air at 1450°C for 2h. Details on the study of aMJ processes on the standard sized bars and disks can be found elsewhere [16].

aDLP1 FDPs were processed using a prototype DLP Printer equipped with a 405 nm wavelength UV light and X/Y plane resolution of 50 μm pixels. Parts were printed with 25 μm layer thickness. After printing the parts were first cleaned with a mixture of 50 vol.% HDDA and 50 vol.% ethanol in an ultrasonic bath to remove any uncured slurry. After debinding the parts were sintered at 1480°C for 2h. No additional postprocessing was performed. Details on the aDLP1 processing on the standard sized bars and disks can be found elsewhere [15]. aDLP2 FDPs were manufactured using a commercial ZrO₂ slurry (LithaCon 3Y 230, Lithoz) with a layer thickness of 25 μm . The printed FDPs were cleaned to remove the excess slurry. All the 3D-printed FDPs were debound and finally sintered for 2h at 1450°C in air. Further information on the other processes can be found in the literature on SLA [17] and DLP [18-21].

Main characteristics of the 79 FDPs samples are summarized in Table 2

Table 1: Main features of four-unit fixed dental prostheses (FDPs) samples

Sample	Technology*	Geometry (4-unit)	Raw material (3Y-TZP)	Number of FDPs (n)	Layer thickness (μm)	Sintering in air ($^{\circ}\text{C}/\text{h}$)
s1	CAM milling		Zirconia GC	16	-	1450/6.5
s2			ST		-	1450/2
aSLA	SLA		Zirconia Pritidenta	16	25	1450/2
aMJ	MJ		Zr-P03 paste	11	10.5	1450/2
			C800 7250001 ink	16		
aDLP1	DLP		3Y-TZP	16	25	1480/2
aDLP2		LithaCon 3Y 230 slurry	4	25	1450/2	

*CAM: Computed assisted manufacturing; SLA: Stereolithography; DLP: Digital light processing; MJ: Material jetting

2.2 Sample digitalization and repeatability testing

Initially, four procedures for the light-optical digitalization were tested for the highest repeatability in preliminary trials to evaluate their feasibility. Due to strong reflections, white and tooth-colored monolithic ceramics are difficult to measure. Coating with powders can improve scannability, but manual application of the powder and the associated nonuniform thickness of the coating can cause inaccuracies. Therefore, a powder-free scan approach was preferred. The four different technologies included a non-contact 3D profilometer with stripe light pattern at 25x magnification (VR-52000, Keyence, Osaka, Japan), a light optical 3D coordinate measuring machine (VL-500, Keyence, Osaka, Japan), a confocal laser scanning (TRIOS 4, 3shape, Copenhagen, Denmark) and a dynamic depth scanning technology (Primescan, DentsplySirona, York, USA). Each FDP was digitized five times repeatedly with each technology. The scans of the different technologies were compared with each other using a 3D target/actual comparison and the repeatability was determined. The optical 3D coordinate measuring machine showed a positive mean deviation of 21 μm and a negative mean deviation of -18 μm . Only 78 % of the data points were within the tolerance of 20 μm . With the non-contact 3D profilometer, due to the stationary positioning during the measurement, data acquisition in undercut areas was difficult. Confocal laser scanning had a mean deviation of ± 12 μm with 95 % of the data points within the tolerance of 20 μm . With a positive mean deviation of 7 μm and negative mean deviation of -6 μm , dynamic depth scanning showed the highest repeatability. Across the five scans, 98% of all data points were within a tolerance of 20 μm . Dynamic depth scanning was therefore used for digitizing the FDPs. All 79 FDPs were digitized 360° by an experienced operator under consistent lighting conditions utilizing a single scan, so that it was not necessary to merge several files. Scan data was then saved as standard triangulation of tessellation language (STL) datasets at maximum resolution.

2.3 Dynamic depth scanning: accuracy measurement procedure

Dynamic depth scanning data were subsequently analyzed with the inspection software Geomagic Control X (Geomagic, 3DSystems) using a three-dimensional target/actual comparison. For this purpose, the planning file was first defined as a target and segmented into the three analysis areas -inner shell, margin and outer shell, as shown in Fig. 1. The margin corresponded to a 0.25 mm horizontal step and 0.3 mm rising transition to the outer shell (magnification in Fig. 1). Subsequently, superimposition was performed with the samples using an initial-fit followed by a best-fit alignment which uses an iterative closest point algorithm. For the best-fit superimposition, areas where support structures were attached, pressure

artefacts and fractured edge areas were excluded from the superimposition. The three-dimensional surface comparison was performed separately for the three segments. The previously excluded areas were also not included in the evaluation.

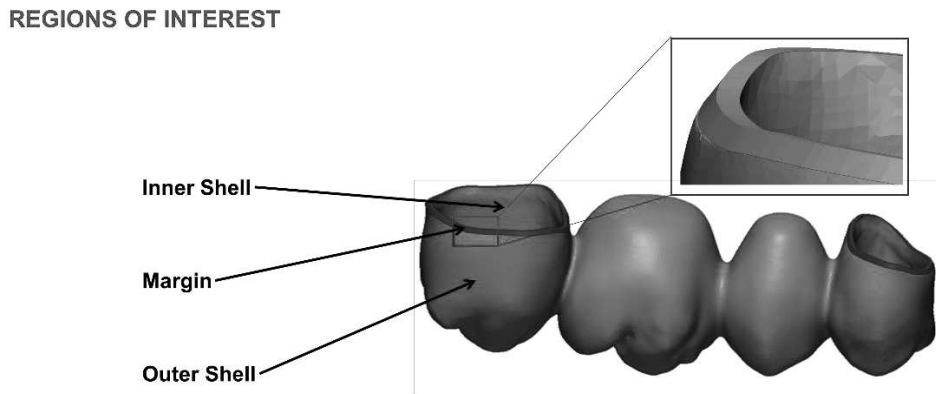


Fig. 1: Image of the regions of interest represented by the inner shell, margin and outer shell of the four-unit bridge.

2.4. Statistical analysis

The statistical evaluation of the surface deviation was performed using the root mean square error (RMSE) method, and the descriptive evaluation of the deviations was performed using the positive and negative mean surface deviation. The data were first tested for normal distribution using Shapiro-Wilk test. Subsequently, a nonparametric Kurskal-Wallis test with post hoc Wilcoxon-Mann-Whitney U test was performed. Significance values have been adjusted by the Bonferroni correction for multiple tests. The statistical analysis was carried out with an established software program (IBM SPSS Statistics, v24.0; IBM Corp) ($\alpha = 0.05$).

3. Results

Fig. 2 shows examples of zirconia FDPs produced according to the STL-file (Fig. 2a) by means of additive (Fig. 2b) and subtractive (Fig. 2c) manufacturing.

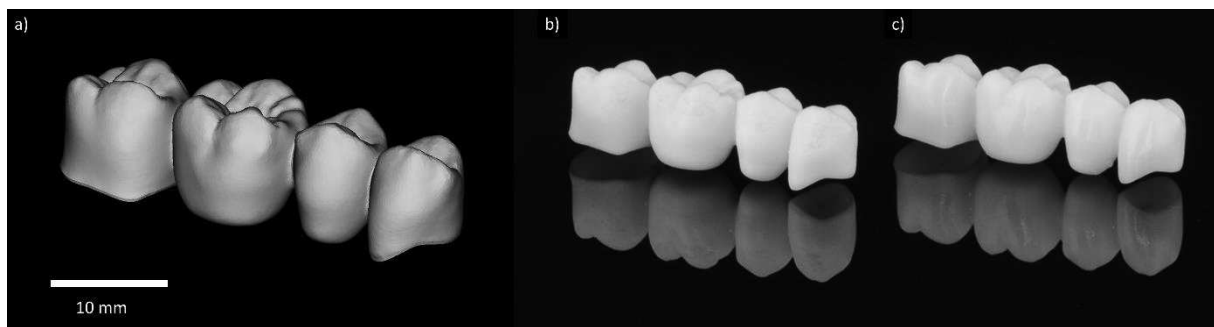


Fig. 2: a) STL of a four-unit zirconia fixed dental prostheses, b) example of an additively manufactured (aMJ) FDP , c) FDP produced by subtractive manufacturing.

Heat maps comparing the measured data with the CAD file of the samples manufactured with the different technologies are shown in Fig. 3.

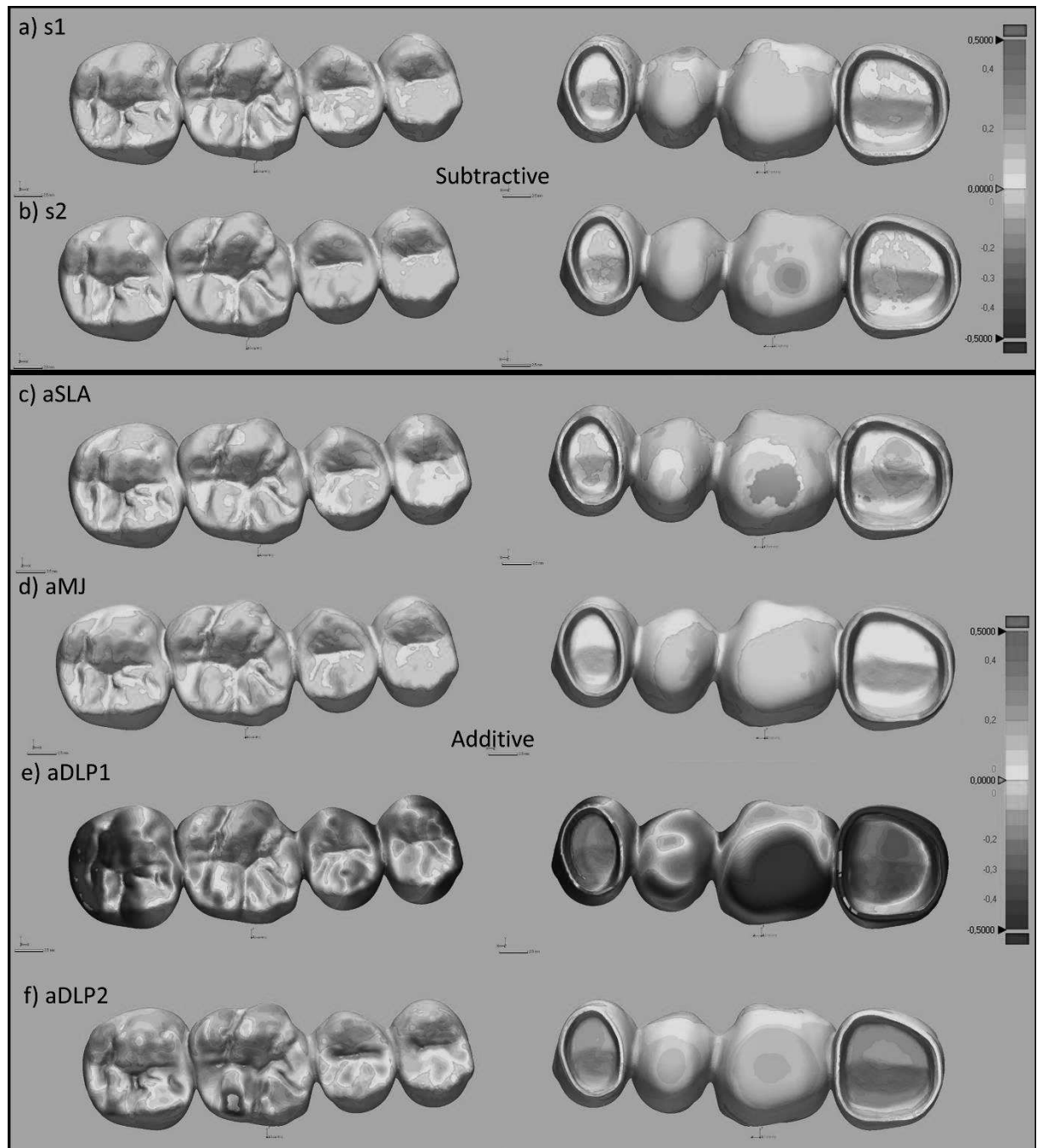


Fig. 3: Representative heat maps visualizing the deviation of the scanned samples compared to the CAD file. a)-b) subtractive (s1 and s2), c) stereolithography (aSLA), d) material jetting (aMJ), e)-f) digital light processing (aDLP1 and aDLP2). Yellow and red colored surfaces represent areas with a positive deviation (excess material). Blue color shows a negative deviation and therefore a lack of material.

The root mean square error of the samples produced with the different manufacturing technologies is shown in Table 2 and visualized in Fig. 4 for the outer shell, the margin, and the inner shell.

Table 2: RMSE of different ROI.

	RMSE (μm)		
	Outer Shell	Inner Shell	Margin
s1	27 ± 4	28 ± 6	27 ± 6
s2	44 ± 7	40 ± 8	49 ± 15
aSLA	73 ± 16	68 ± 8	70 ± 46
aMJ	46 ± 17	53 ± 19	125 ± 109
aDLP1	243 ± 26	288 ± 48	389 ± 80
aDLP2	108 ± 15	141 ± 7	67 ± 40

Detailed positive and negative mean surface deviations are presented in Fig. 5 and Fig. 6.

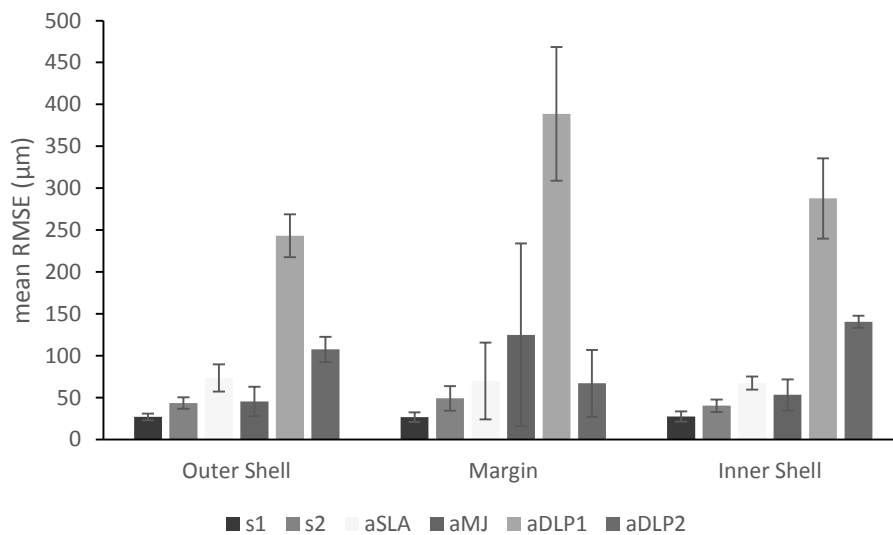


Fig. 4: Root mean square error (RMSE) of the samples produced with different manufacturing technologies for the outer shell, margin and inner shell.

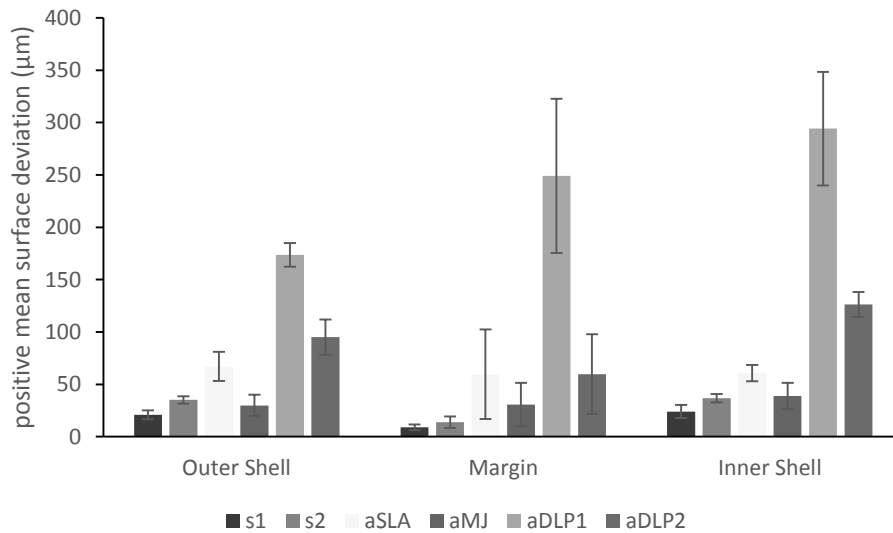


Fig. 5: Positive mean surface deviation from the STL file of the samples produced with different manufacturing technologies for the outer shell, margin and inner shell.

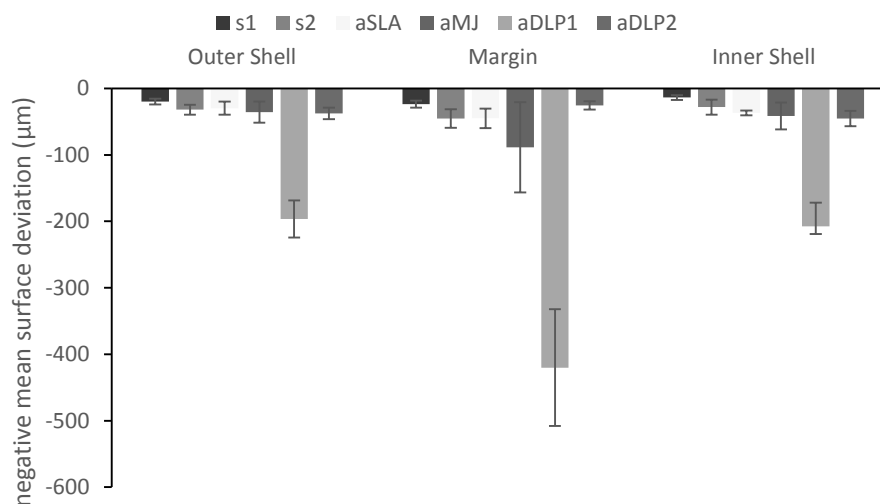


Fig. 6: Negative mean surface deviation from the STL file of the samples produced with different manufacturing technologies for the outer shell, margin and inner shell.

3.1. Inner Shell

Kruskal-Wallis test revealed significant differences in RMSE between groups ($P < 0.001$). The smallest differences were shown by s1 with no significant difference from s2 ($P = 0.297$). Compared with s2, aMJ ($P > 0.999$) and aSLA ($P = 0.085$) showed no statistically significant differences. aDLP1 showed significantly higher deviations than the other groups ($P < 0.001$). The negative mean deviations of all groups were smaller than the positive mean deviations. Only aDLP2 and aDLP1 showed mean surface deviation $> 100 \mu\text{m}$. All others showed mean

deviations of $< 50 \mu\text{m}$ for the positive and negative mean surface deviation, except for aSLA with $< 100 \mu\text{m}$ for the positive mean surface deviation.

3.2. Margin

RMSE shows significant differences between the groups ($P < 0.001$). Congruent with inner shell, s1 showed the least deviation with no significant difference from s2 ($P = 0.097$), and s2 did not differ from aMJ ($P = 0.274$) and aSLA ($P > 0.999$). The highest deviations were shown by aDLP1 ($P \leq 0.080$). The pMean was $< 50 \mu\text{m}$ for s1, s2, and aMJ and $< 100 \mu\text{m}$ for aSLA and aDLP2. The negative mean surface deviation was $< 50 \mu\text{m}$ for all groups. Only aDLP1 showed higher mean surface deviations.

3.3. Outer shell

RMSE shows significant differences between the groups ($P < 0.001$). The smallest deviations were shown by group s1 without significant difference to group s2 ($P = 0.054$) and aMJ ($P = 0.121$). aSLA showed larger deviations that did not reach the significance level to s2 and aMJ ($P \geq 0.103$). Again, aDLP1 showed the largest deviations. The positive mean surface deviation was $< 50 \mu\text{m}$ for s1, s2, aMJ, and $< 100 \mu\text{m}$ for aSLA and aDLP2. The negative mean surface deviation was $< 50 \mu\text{m}$ for all groups. Only aDLP1 showed higher mean surface deviations.

4. Discussion

4.1. Main findings

In this study, the accuracy of zirconia four-unit FDPs produced by additive manufacturing was analyzed. Subtractively manufactured FDPs (s1 and s2) served as reference samples. Additive manufacturing included three different vat photopolymerization processes (aSLA, aDLP1 and aDLP2) and one material jetting technology (aMJ). The analysis was conducted as a three-dimensional target/actual comparison based on the RMSE segmented into the three areas: inner shell, margin, and outer shell. The two subtractive processes showed the lowest deviations, followed by the MJ process and the SLA process. aDLP2 was only described descriptively due to the low number of samples but showed larger deviations than the previous groups. Samples of the group aDLP1 had the highest deviations, nearly twice as large as the other groups. This indicates that, regarding the accuracy, state-of-the-art subtractive manufacturing processes for fixed dental prostheses can still be considered superior to additive manufacturing.

4.2. Methodology

4.2.1. Sample preparation

Dental ceramics made from 3Y-TZP are currently mostly produced by CAD/CAM milling or grinding depending on the sintering stage [22]. This technique has been continuously improved over the past decades and provides reliable, strong FDPs [23]. Even though the subtractive manufacturing route is currently most used, AM of dental ceramics is drawing a lot of interest. Ceramic AM companies have started developing SLA and DLP printers focused on printing ceramics for dental applications. SLA and DLP has been used to accurately manufacture dental crowns and bridges [17, 18, 24-26]. Another popular and highly accurate AM technique for producing 3Y-TZP is material jetting, also known as inkjet printing. MJ allows to build up highly dense and accurate green bodies, speeding up the postprocessing route due to the limited amount of binder that has to be removed compared to SLA [27]. High strength 3Y-TZP ceramics have been reported to be obtained by MJ when printing in the preferable 0° direction [16, 28-30]. 3Y-TZP 3D-printed dental crowns have been produced, demonstrating the possibility to create accurate restorations using material jetting [29].

4.2.2. Measurement method

Studies on the accuracy of manufactured FDPs can be divided into two-dimensional and three-dimensional methods. In two-dimensional measurements, distances between two lines are measured. This can be done non-destructively by an impression replica technique using a silicone material and subsequently measuring the cross-sections, or destructively by cementing the FDPs, forming cross-sections and measuring the width of the cement gap afterwards. Three-dimensional methods include the measurement of the cement gap after taking X-Ray Tomography (μ CT) images of the FDPs and using three-dimensional target/actual comparisons. Compared to measurements of the cement gap, the latter has the disadvantage that a clinically relevant fit is not determined. On the other hand, it is advantageous that the entire object surface is analyzed and three-dimensional shrinkage and distortion can be displayed by means of heat maps which is suitable for exploratory analyses.

Sample digitalization methods determine the methodical error in digital accuracy analysis. The gold standard was a long time represented by coordinate measuring machines based on tactile measuring technologies while nowadays also optical technologies are comparable with the advantage of 360° measurements. The use of an optical 3D coordinate measuring machine and a 3D profilometer were evaluated in preliminary tests, but intraoral scanning using the dynamic depth scanning technology showed the highest repeatability. This can be explained by the possibility of non-stop 360° scanning of the object, when using the mobile scanner. Using the

other mentioned technologies for optical scanning, separate data sets of the inner and outer shells needed to be assembled by the software, resulting in a negative impact on repeatability. A three-dimensional surface comparison was chosen as the method for analysis. This allows the entire object surface to be analyzed and the level of deviations to be visualized as heat maps. This method is suitable for comparing relative differences between groups, but needs to be considered as an exploratory analysis. Comparable studies to this work did not evaluate the accuracy of the FDPs but derived it from the fit in predominantly two-dimensional measurements [17]. This was mostly evaluated by relining with silicone materials and sectioning or destructive analysis based on the width of the cement gap in cross-sections. An alternative would have been the digitization method using a μ CT. However, the mean surface deviations of $< 7 \mu\text{m}$ on repeated scans were considered clinically sufficient.

A risk of bias of the present study might be seen in the manual exclusion of artifacts and support structures during evaluation. Since the focus was on warpage and shrinkage, these were excluded so that isolated partially operator-dependent areas would not affect the results.

4.3. Subtractive vs additive manufacturing technologies in relation to clinical needs

CAD-CAM fabrication of multi-span fixed restorations has been extensively studied, still the standard for a minimum accuracy requirement in ceramic dentistry is not yet defined [31]. In a systematic review, Hasanzade et al. [32] investigated digital and conventional fabrication methods for the fabrication of FDPs, with the tendency for the best internal and marginal fit being achieved with a fully digital workflow. However, this included only subtractive procedures. The clinical relevance of the deviations depends on the region and its influence on the fit. It is generally accepted that a marginal gap width smaller than $120 \mu\text{m}$ is clinically acceptable [33]. Overall in the entire internal and marginal fit of the FDP, a maximal deviation of $100 \mu\text{m}$ is reported as the limit to be considered clinically acceptable [34, 35]. The external accuracy is influenced by final machining and polishing, but should be as low as possible in a model-free production in order not to negatively influence the occlusion. The final fit of the FDP will be determined by its accuracy, which depends strongly on the fabrication process [17, 18, 26]. Poor accuracy and the concomitant poor fit result in a higher risk of plaque accumulation, inflammation and reduced strength of the FDP [36, 37].

The two subtractively fabricated groups (s1 and s2) showed the lowest deviations in this study with mean deviations smaller than $50 \mu\text{m}$ in all areas, which is in accordance with earlier studies that reported marginal discrepancy of 17 to $118 \mu\text{m}$ [26, 38-52] for milled zirconia bridges.

Furthermore, there is no significant difference between the FDPs s1 (GC) and s2 (Pritidenta) ($p > 0.096$). Both these results indicate the high quality of the subtractively produced FDPs.

The additive manufactured groups showed deviations in accordance to the maturity degree of the technology. While the samples produced by commercially available systems (aSLA, aMJ, aDLP2) had mean negative deviations of $-88 \mu\text{m}$ and mean positive deviations $+126 \mu\text{m}$, the prototype DLP-Printer (aDLP1) revealed the highest negative mean deviation of $-420 \mu\text{m}$ and positive mean deviation of $+294 \mu\text{m}$. Although high-strength standard shaped disk and bars can be manufactured by the identical route used by aDLP1, the fabrication process and shrinkage adjustment to make complex shaped bridges needs to be tailored.

Several studies analyzed the trueness of vat polymerization (SLA and DLP) produced molar crowns, reporting that it was possible to produce crowns with highly accurate dimensions, but that the gap width was larger than what can be obtained by subtractive manufacturing and possibly not sufficient yet for clinical application [17]. The average deviation of a DLP printed implant was reported as $89 \mu\text{m}$ and $-129 \mu\text{m}$ ($\pm 68 \mu\text{m}$) homogeneously distributed along the length of the printed implant [53].

Alumina single crowns manufactured using aDLP2 were reported to show a dimensional accuracy of $41 \pm 11 \mu\text{m}$ and zirconia single crowns manufactured by a SLA process had an accuracy of $65 \pm 6 \mu\text{m}$ [18]. Both AM produced crowns had a higher accuracy than subtractive milled zirconia crowns ($72 \pm 13 \mu\text{m}$) which is explained by the higher precision of the laser and photodiode compared to subtractive methods [18]. The accuracy of the single crowns produced by vat polymerization was considered within clinically acceptable limits [18, 26].

One study reported most defects on the inner and the marginal area [17], whereas another reported occlusal grooves showed the highest positive deviations, caused by the layerwise buildup, having the highest influence on curved surfaces [26]. While it was possible to observe the layerwise buildup on the surface of our samples, similar to other studies [17, 18, 26], only minor differences in accuracy between outer shell, inner shell and margin were observed.

Multiple parameters influence the dimensional accuracy during manufacturing including layer thickness, degree of polymerization, excessive polymerization, polymerization and sintering shrinkage and staircase effect [54, 55]. In contrast to additive manufacturing of polymer materials, the accuracy of AM ceramics is not only affected by the printing technique, but also by the factors chosen for the manufacturing technique, like slurry composition, layer thickness, polymerization parameters as well as by the following downstream process steps such as

cleaning, debinding and the shrinkage of the ceramic during sintering [18, 55]. When analyzing the accuracy of a DLP produced FDP green body, it was reported that both pits and protrusions resulted in the highest deviations due to support damage [24]. According to the authors, negative deviations at the axial surface and margin area could have been caused by inadequate compensation for sintering shrinkage [26]. Less deviation and shrinkage in restorations manufactured with DLP is reached for short-unit restorations [56]. Especially for 3D-printed ceramics, controlling the shrinkage and warping during sintering is more challenging for FDPs than for smaller parts like single crowns. For subtractively produced single crowns, RMSE of $52 \pm 18 \mu\text{m}$ (outer shell), $43 \pm 12 \mu\text{m}$ (inner shell) and $35 \pm 7 \mu\text{m}$ (margin area) compared to a RMSE of crowns produced with aSLA of $53 \pm 9 \mu\text{m}$ (outer shell), $38 \pm 12 \mu\text{m}$ (inner shell) and $34 \pm 5 \mu\text{m}$ (margin area) were reported [26]. For our subtractively manufactured four-unit bridges s1, RMSE of $27 \pm 4 \mu\text{m}$ (outer shell), $28 \pm 6 \mu\text{m}$ (inner shell) and $27 \pm 6 \mu\text{m}$ (margin area) were estimated. For FDPs s2, RMSE of $44 \pm 7 \mu\text{m}$ (outer shell), $40 \pm 8 \mu\text{m}$ (inner shell) and $49 \pm 15 \mu\text{m}$ (margin area) was measured. The higher RSME found in our aSLA FDPs of $73 \pm 16 \mu\text{m}$ (outer shell), $68 \pm 8 \mu\text{m}$ (inner shell) and $69 \pm 46 \mu\text{m}$ (margin area) shows the challenge for processing highly accurate long-span dental restorations, like the investigated four-unit FDPs.

Subtractive manufacturing uses prefabricated blanks which ensure a homogenous powder compact. This leads to a controllable shrinkage and high predictability of the FDP dimensional changes during sintering. On the other hand, the homogeneity of the printed green body can be disturbed, e.g. by the layered structure leading to anisotropic shrinkage during sintering [20, 57]. This can negatively influence the predictability and the final accuracy of the 3D-printed components.

aMJ has shown the best results among the studied AM FDPs for the inner and outer shell, with results close to the subtractive manufactured FDPs and no significant differences to s2. Due to the lower layer thickness ($10.5 \mu\text{m}$ compared to $25 \mu\text{m}$) and the high green body density, this technique allows for the production of parts showing trueness comparable to conventional manufacturing, within the clinically acceptable limit of $<100 \mu\text{m}$. The margin area of aMJ FDPs shows a higher deviation because the thin restoration margins tended to break during the support removal in water, leading to higher negative deviations. It has been reported that it is possible to acquire highly accurate designs using the XJET printer, but that the printing of complex designs with fine features, like a thin restoration margin, are still challenging due to the support removal [16, 58].

4.4. Challenges to be overcome and perspectives

Subtractive manufacturing in dentistry has evolved over the past decades into a mature technology resulting in precisely adjusted manufacturing systems and processing parameters. AM of ceramics for dental applications is an emerging technology combining many processing parameters that still need to be tailored. Especially the control of shrinkage behavior is important to obtain accurate final restorations [17]. Intensive research on improving the additive manufacturing methods, devices and material is still necessary to obtain high quality dental materials [59].

At the moment, AM of ceramics is a time-demanding manufacturing process. Grinding of the s1 bridges was performed in 30 min. Sintering of milled restorations takes, according to manufacturer recommendation approx. 7 h, but can be reduced to approx. 90 min with state of the art high-speed sintering cycle proposed for single crowns and 3-unit bridge restorations [60, 61]. New research shows that high-speed sintering in just 30 min, including cooling time, is possible [62]. However, 3D printing is much more time demanding. In systems like DLP or SLA, the polymer matrix needs to be thermally removed in a debinding step. The steps of drying and debinding can take more than 20 hours for smaller cylindrical parts with 4 mm in diameter [63]. In this work, more than four days of debinding were applied to aSLA FDPs. Debinding and sintering of alumina crowns fabricated with aDLP2 was reported to take 200 h [18].

Furthermore, the ceramic green bodies produced by AM are often handled manually during subsequent post-processing, which has a higher risk of introducing new defects, which also increases the cost of the additive manufacturing process [19, 21]. Therefore, further development is needed to reach the productivity of modern subtractive manufacturing systems and additional investigations for the reliability of the produced parts for long-term use are necessary.

5. Conclusion

Significant differences were calculated between the different subtractive and additive manufacturing technologies ($p < 0.001$) used for preparing zirconia FDPs. The two subtractive groups were the most accurate with no significant difference regarding the used material or device ($s1/s2$, $p > 0.054$). Likewise, no statistical difference regarding accuracy was found when

comparing the subtractive group s2 with the additive manufactured FDPs, aMJ and aSLA, in most ROIs ($p>0.085$). In general, mean surface deviation was $<50\ \mu\text{m}$ for s1, s2 and aMJ and $<100\ \mu\text{m}$ for aSLA and aDLP2. aDLP1 showed surface deviations $>100\ \mu\text{m}$ and was the least accurate method, compared to the other additive/subtractive technologies.

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