3D Printed Materials Dentistry

Editor

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About the Editor

Kathrin Becker

Kathrin Becker studied Dentistry at the University of Greifswald and Gottingen (Germany), as well as Applied Computer Science (Bachelor and Master of Science) at the University of Gottingen. Her master's degree was awarded with distinction, and she was awarded with a short-term research scholarship at the University of Pittsburgh (USA). She received her Doctorate (magna cum laude) from the University of Dusseldorf (2014). She performed her specialty training in orthodontics from 2014–2018 at the same university under the leadership of Prof. Dr. D. Drescher. She completed a Habilitation in Dentistry and Maxillofacial Medicine in 2019. Since 2021, she has been the Assistant Medical Director at the Department for Orthodontics (University Hospital Dusseldorf). Currently, she is member in various committees from international societies (Osteology expert council, EAO congress committee, EAO junior committee, and Osteology Scientific Review Board), Editorial Board member (*Journal of Clinical Periodontology*, *Clinical Oral Implants Research*, and *Clinical Oral Investigations*), and Editor-in-Chief of the Wiley and Sons Ltd. *Journal Clinical and Experimental Dental Research*. She has published 90 peer reviewed articles and has been awarded with research prizes: Osteology (2019), Young Scientists Award (2014), and Publons Reviewer Award (1× 2019, 2× 2020), and she has been selected by the University of Dusseldorf for the Selma Meyer Mentoring Programme and the Academic Career Development Programme. Her research interests comprise but are not limited to skeletal anchorage, application of 3D-imging in dentistry, e-learning in dentistry, and micro-computed tomography.

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Editorial

3D-Printed Materials Dentistry

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This editorial focuses on the Special Issue on 3D-printed materials in dentistry. The articles of this Special Issue cover a wide range of applications in the dental field. They range from applications in the clinical workplace to in vitro investigations on the impact of postcuring and storage time on accuracy and precision measurements to the indirect transfer of orthodontic brackets, as well as the application of 3D printing in dental education. Five out of eleven research papers deal with applications of 3D printing in orthodontics, one study presents a 3D-printed fitting system for FFP2 masks which were applied during the COVID-19 pandemic, while the remaining one addresses applications in prosthodontics and restorative dentistry.

In orthodontics, 3D printing is becoming increasingly popular. It allows to create individualized appliances, to perform different treatment tasks at the same time, and much more [1].

The article by Küffer et al. [2] demonstrates a digital workflow for producing highly individualized, skeletally borne 3D-designed appliances for the upper and the lower jaw. As orthodontic mini-implants providing skeletal anchorage are often inserted using templates, the article also presents typical designs for insertion guides. Additionally, it highlights potential sources of errors within the digital workflow that are especially relevant to clinicians in their daily practice.

The article by Kirschner et al. [3] addresses the question of whether steam autoclaving impacts on the biomechanical properties of 3D-printed insertion guides. Autoclaving is necessary as the guides may be in contact with blood during orthodontic implant placements. The study compared two autoclaving cycles and different resin/printer combinations. For the protocol using a lower temperature and a longer autoclaving time, no biomechanical alterations could be observed, whereas in some groups, significant difference were noted for the faster autoclaving protocol at an increased temperature.

The article by Ihssen et al. [4] investigated whether the socket height of 3D-printed casts used to thermoform aligners has an impact on the aligner thickness and homogeneity. Thicker aligners would apply higher forces to the teeth, and varying thickness values might lead to inhomogeneous force applications across the aligners. Indeed, the study demonstrated that increased socket height was associated with thinner and more homogeneous aligners. Additionally, in all groups, the thickness values were highest at the incisal surfaces, and they were the lowest at the facial aspects, especially at the cervical margins. Future clinical studies are needed to assess the impact of local variation in the aligner thickness on the local force application and thus on treatment's success.

The article by Jungbauer et al. [5] compared two 3D-printed bracket transfer trays with different shore hardnesses and assessed whether the crowding of the incisors impacts on the accuracy of indirect bracket transfer. Additionally, the workflow was tested using two different methods, i.e., intraoral scanning and Micro-CT, whereby intraoral scanning was inferior as the shape of the brackets was very different from that of the reference. The study found minor linear deviations for the indirect bracket transfer, whereas angular deviations and deviations in the torque reached values that were above the limits specified by the American Board of Orthodontists. Finally, the deviations were more pronounced for crowded teeth.

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Mieszala et al. [6] proposed the fabrication of Polymer Polyether Ether Ketone (PEEK) transpalatal arches designed in a CAD process. The forces and moments of two transpalatal arch designs were within a range that appears to be clinically useful for expansion or for anchorage purposes. However, their clinical applicability is still limited as the PEEK arches cannot be activated intraorally. Furthermore, there is also a lack of clinical studies proving their applicability in clinical settings.

Vichi et al. [7] focused on infection control during COVID-19 pandemic. They designed a mask fitter to ensure proper fit of FFP2 masks. While no data are available on the efficacy of this device, they reported a high satisfaction of staff members with the novel mask fitting system.

The article focusing on dental education by Lugassy et al. [8] described the fabrication of multi-colored teeth to train students in Class I preparations of molars. The study demonstrated low reliability in evaluations of undergraduate students given by staff for conventional plastic molars, whereas moderate to good reliability was found for the multicolored 3D-printed teeth. Thus, it was concluded that the multicolored teeth can provide a more objective evaluation of the students' performance in cavity preparation.

The article by Lin et al. [9] focused on the dimensional stability of 3D-printed casts obtained using digital light processing (DLP) or stereolithography (SLA). While no shrinkage was noted in the first two weeks for DLP and SLA groups, a significant contraction was noted from two to six weeks.

The article of Doh et al. [10] focused on the dimensional changes during the postcuring of 3D-printed denture bases. They found that the dimensional changes increased with postcuring times of 15–60 min, and that accuracy was higher when no prior removal of support structures had been performed.

The article by Lüchtenborg et al. [11] outlined the potential applications of fused filament fabrication (FFF) printing in dentistry. Despite FFF was reported to exhibit a reduced accuracy compared to that of other 3D printing technologies, this technology was reported to be a promising and cost-efficient alternative for the in-house production of digitally designed models, trays, or prototypes for denture try-ins.

The article by Çakmak et al. [12] investigated the impact of the 3D printing layer thickness on the trueness and margin quality of interim dental crowns in comparison to that of milled PMMA crowns. They demonstrated that milled PMMA crowns had the highest margin quality, while printed crowns with a layer thickness of 20µm and 100µm had the lowest. They also found that the trueness and marginal quality of the 3D-printed interim crowns were influenced by the printing layer thickness. Trueness of milled crowns was reported to be superior compared to 100 µm printed crowns, and margin quality was also highest for milled crowns.

In summary, the publications of this Special Issue reflect the multitude of areas in which 3D printing technologies can be applied to dentistry at present, and they also outline future avenues for novel research projects. As the majority of studies were performed in vitro, this Special Issue might also stimulate future clinical studies and systematic reviews of the existing literature.

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Communication

Application of the Digital Workflow in Orofacial Orthopedics and Orthodontics: Printed Appliances with Skeletal Anchorage

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Abstract: As digital workflows are gaining popularity, novel treatment options have also arisen in orthodontics. By using selective laser melting (SLM), highly customized 3D-printed appliances can be manufactured and combined with preformed components. When combined with temporary anchorage devices (TADs), the advantages of the two approaches can be merged, which might improve treatment efficacy, versatility, and patient comfort. This article summarizes state-of-the-art technologies and digital workflows to design and install 3D-printed skeletally anchored orthodontic appliances. The advantages and disadvantages of digital workflows are critically discussed, and examples for the clinical application of mini-implant and mini-plate borne appliances are demonstrated.

Keywords: digital workflow; orthodontic; skeletal anchorage; temporary anchorage device; 3D printing; printed appliance; metal printing

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1. Introduction

Digital technologies, such as 3D scanning and printing, have expanded the range of treatment options in various medical disciplines in recent years [1–5]. In dentistry, a wide range of possible implementations have been reported across specialties, including the fields of orofacial orthopedics and orthodontics [6]. In contemporary orthodontics, scanning dental arches for metric analyses or the creation of digital set-ups are routinely performed. A relatively new application is the design and clinical application of individual metal-printed orthodontic appliances [7]. Especially in complex cases, where conventional techniques and preformed devices do not fulfill all requirements, individualized 3D-designed and printed appliances may be advantageous [7,8]. In addition, appliances created using a digital workflow are reported to offer many advantages in terms of patient comfort, treatment efficacy, and predictability [7,8].

As every orthodontic (and orthopedic) force is associated with a reactive force of equal magnitude, orthodontic treatment success necessitates sufficient anchorage to avoid side effects, including undesired tooth movement of dental anchorage units [9,10]. Therefore, orthodontic anchorage is a term for all measures preventing those reactive forces and moments [11]. Typically, either teeth (dental units) or extraoral attachments are used. Extra-anchorage can be achieved by integrating the bone into the resistance unit by means of so-called temporary anchorage devices (TADs). This concept is called skeletal anchorage, and it gained popularity in recent years owing to a reduction in side effects and increased treatment efficacy [8–10,12–17]. Among the TADs, orthodontic mini-implants are frequently employed due to their ease of application and the favorable cost–benefit ratio [18,19].

In the upper jaw, the anterior palate proved to be the most favorable insertion site with the highest survival rates [15,20,21]. In the mandible, however, mini-implants usually have to be inserted into the alveolar ridge [22–24], where failure rates range from 14% to 16% [25–27]. The lower success rates were mainly attributed to the reduced bone quality and the proximity to the dental roots. In addition, a risk of root contact or even penetration exists [25,26]. Since mini-implants are inserted between the roots, the extent of possible orthodontic tooth movement is implicitly limited. Mini-implants inserted into the mandibular buccal shelf did not gain much acceptance in most of the countries because of their very low survival rates [27], despite their advantage not to limit tooth movements. Alternatively, the chin below the dental roots was identified as a possible insertion site due to its high bone quantity and quality. Nonetheless, mini-implants are not well-suited for the area, and thus special mini-plates, such as the Mentoplate, were developed [28,29]. These plates can be fixed to the bone using regular osteosynthesis screws, but require a slightly more invasive surgical procedure for insertion and removal, when compared with mini-implants.

TAD-borne appliances are regarded as “non-compliance” devices as patients only need to keep them clean, but are not requested to wear them for a certain amount of hours (they have to wear them 24 h/day) [30]. These appliances also permit longer time intervals between appointments and hence a reduction of the overall chair-time [10,12,31].

The application of digital workflows to design and manufacture TAD-borne appliances appears to be particularly promising. The placement of the mini-implants can be planned, which allows the identification of the optimal insertion position based on digital models and, optionally, using radiographs. Utilizing printed insertion guides [32] or the appliance itself [33] allows for implant insertion and appliance installation in one appointment. Owing to the precise fitting, the risk of screw overloading, and therefore the risk of implant failure, can be reduced [34,35]. However, digital workflows rely on the skills and experience of the operator, and pitfalls exist that can limit the advantages of the digital workflow.

The present article describes a fully digital workflow for the creation of TAD-borne metal printed appliances in orthodontics and orofacial orthopedics. Accordingly, some of the most common printing techniques and materials, possible advantages and disadvantages, and potential sources of error are critically discussed.

2. Digital Workflow: From Intraoral Scan to the Printable Data

2.1. Data Acquisition

A fundamental basis for the digital design of orthodontic appliances is gathering information regarding the intraoral situation including teeth, alveolar ridge, and soft tissues. This can be achieved using intraoral scanning or by employing digitized conventional plaster casts. Some studies found the intraoral scan to be more time-consuming than conventional methods [36,37], whereas others did not observe any significant difference [38]. This might indicate that the practitioners’ level of experience could have a major impact on time efficacy. In addition, several studies underlined a sufficient accuracy and precision of intraoral scans for orthodontic purposes [36–39]. Further advantages of digital impressions compared to conventional ones include gains in comfort for the patient and a reduced amount of laboratory waste [7,8,37,39].

Data can be stored in various formats. Nonetheless, the Standard Tessellation Language (STL) format was shown to be the smallest common denominator for 3D data exchange between devices and software programs [8,36].

2.2. Digital Appliance Design

In order to design and manufacture an appliance, the intraoral scan generally requires further processing, including mesh repair, hole filling, and removal of invalid data. Usually, a base is added to the dental arch to form a model (Figure 1).

Software tools that enable digital planning range from open-source over freeware to proprietary solutions. Currently, there is great variability in software complexity, sometimes necessitating a steep learning curve.

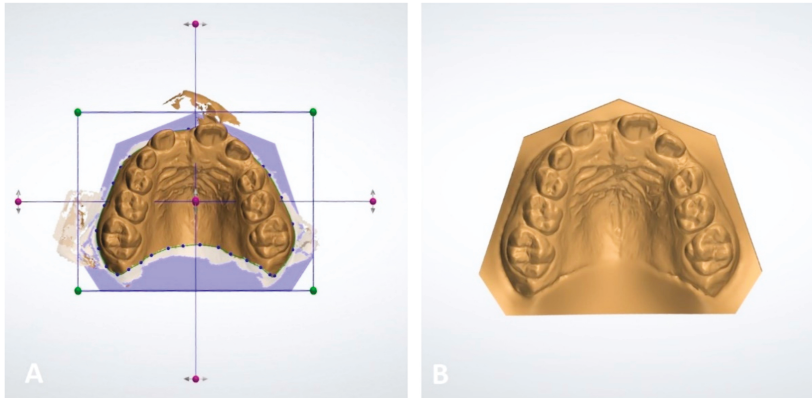

Figure 1. Processing of an intraoral scan to a digital model: (**A**) Removal of noise at the edges, and virtual placement within a digital socket. (**B**) Completed digital model (software: Appliance Designer, 3Shape).

The computer-aided design of the orthodontic appliances enables the planning of all steps, including the positioning of orthodontic mini-implants (Figure 2) or osteosynthesis screws and the position of shells or attachments. Moreover, functional elements can be adapted, for example, by modifying their shape, adjusting the inclination and tilt of gliding mechanisms, or manipulating the expansion of orthodontic screws. Simulation of planned tooth movements is also possible, enabling clinicians to oversee and predict treatment outcomes. Additionally, communication with technicians is simplified. Groups of teeth required to move in a specific direction can be connected to active elements of the appliance, and comparison of actual with predicted tooth movements enables validation (Figure 3).

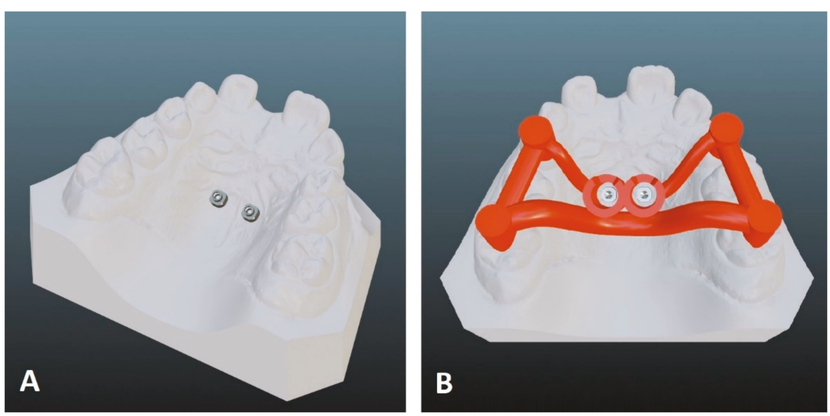

Figure 2. (**A**) Virtual implant planning on a digital model. (**B**) Virtual design of an insertion guide, that can be 3D-printed (software: Blender, Blender Foundation).

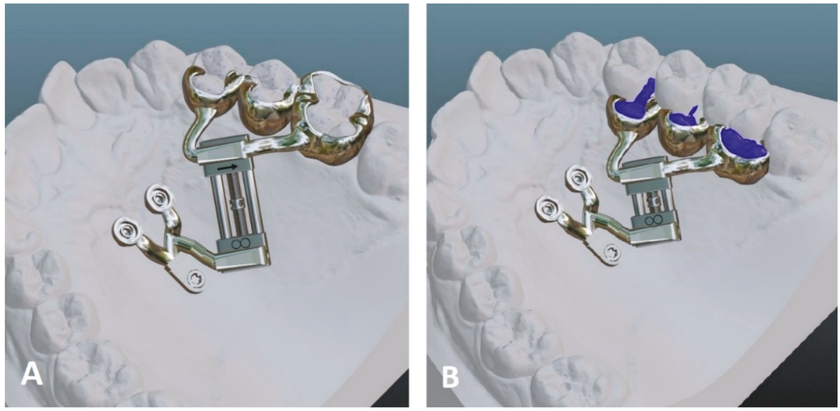

Figure 3. (**A**) Customized active element connected to implants and teeth. (**B**) Simulated activation with estimated movement of teeth (blue) (software: Blender, Blender Foundation).

When skeletal anchorage is employed, an accurate fit of the digitally planned appliance is fundamental to avoid overloading of the orthodontic implants. One option is to perform data acquisition after implant insertion. Sometimes, scanners show problems recognizing the metallic implants, which can be improved using scanning spray or scan bodies [40]. The other option is to insert the mini-implants upon digital planning through guides and utilize the planned position for the appliance design. Digitally planned insertion guides allow implant and appliance insertion within the same visit [40–43] (Figure 2). This can also be achieved by using so-called *Direct Screws*, which are inserted after the adhesive bond of the appliance onto the teeth [33]. As the appliances themselves serve as an insertion aid, no additional guide is needed. When installing a printed *Mentoplate* with supra-constructions for orthodontic tooth movement, a guided insertion is mandatory to precisely achieve the planned position and ensure the fit of supra-construction and teeth.

Digital positioning of (mini-)implants is an already established process in prosthodontic and orthodontic treatments. For insertion of orthodontic screws in the anterior palate, 3D radiographic images are oftentimes not needed. Nonetheless, radiographs can help to estimate the available bone supply [40,44], and they are indicated in case of palatially displaced teeth, limited bone supply, or pathologies, including clefts (Figure 4). Lateral cephalograms provided a sufficient approximation when compared to cone-beam computed tomography (CBCT) for most of the patients [41,45,46]. In case of displaced teeth, patients with craniofacial anomalies, or in case of insufficient bone height visible on a cephalogram, optimal insertion sites and angles can be identified with the help of 3D radiographs [46]. In case of impacted teeth, the optimal vector for force application may also be incorporated in the digital design based on the information from CBCT [12] (Figure 4B).

The digital design of mini-plates for the chin bone requires a CBCT to enable an accurate fit. The segmented chin may be matched with an intraoral scan to create an appliance with transmucosal extensions. As root localization is clearly visible in the CBCT, damage of (unerupted) teeth or roots during screw insertion can be avoided. It is also possible to shape the surface of the bottom of the device in perfect match with the superficial structure of the bone, ensuring wide and evenly flat contact that might prevent side effects.

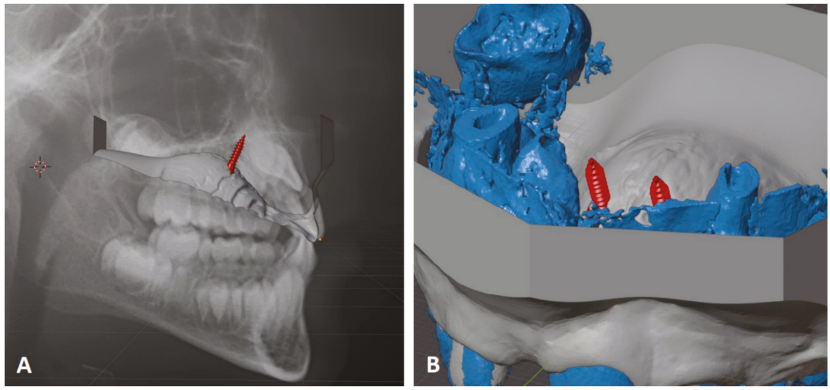

Figure 4. (**A**) Superimposed lateral ceph radiograph for the planning of implant placement. (**B**) CBCT can aid implant placement planning in more complex cases (software: Blender, Blender Foundation).

If clinicians do not design the appliance themselves, the manufacturing laboratory can send the digital draft of the designed appliance back to the clinician, enabling an easy and fast way to revise it, approve it, and communicate desired changes [7,8,47]. Whereas conventionally manufactured appliances would have to be reproduced in case of breakage or improper fit, digital appliances can be reprinted based on the stored data [7], unless they have already been activated in a non-reproducible way.

2.3. Data Preparation for 3D Printing

Since the printing of three-dimensional structures is achieved by adding and joining two-dimensional layers, it is possible to construct very complex and individual structures [48,49]. Printing potential has very few limitations, making preformed pieces just under certain indications necessary. Small parts with internal hollow spaces and pieces with very high requirements regarding the accuracy of fit, such as tubes, elements with internal mechanics, or orthodontic screws, cannot be printed, but can be added by welding them to the printed framework of the appliance. The possibility to create an appliance adapted to the specific intraoral conditions of a patient, almost independent from standardized preformed parts, also allows manufacturing of customized devices for complex and challenging treatment needs [7,47].

The materials used to print the insertion guides for orthodontic implants have to be biocompatible and sterilizable [40]. A precise fit of the insertion tool within the guide's sleeves is needed to achieve an accurate insertion of the implants. Only minor discrepancies, especially in the vertical direction, are tolerable [41]. Vertical control can be improved by using insertion instruments facilitating vertical control [40].

3. 3D Printing Technologies for Orthodontic Appliances

Orthodontic appliances with skeletal anchorage require a metal framework and are usually manufactured by selective laser melting and sintering (SLM/SLS). This additive manufacturing is a kind of solid freeform fabrication and belongs to the powder bed fusion techniques. Since the invention of selective laser sintering (SLS) in 1989 by Carl Deckard, there has been constant progress in the further development of this technique. Using different sources of energy and powdered materials, SLS and SLM can be applied to almost every material melted by laser radiation which solidifies while cooling down, thus making it one of the most versatile additive manufacturing techniques [48,50]. In this technique, a laser emitter moves over a bed of metal powder (e.g., CrCoW) which is heated minimally below its melting point. By tracing the modulated laser over the compacted powder, heat-fusible materials selectively melt in layers. Upon completion of a layer, the powder bed containing the structures being printed is lowered and new material is provided by rolling a thin coating of powder on top (Figure 5).

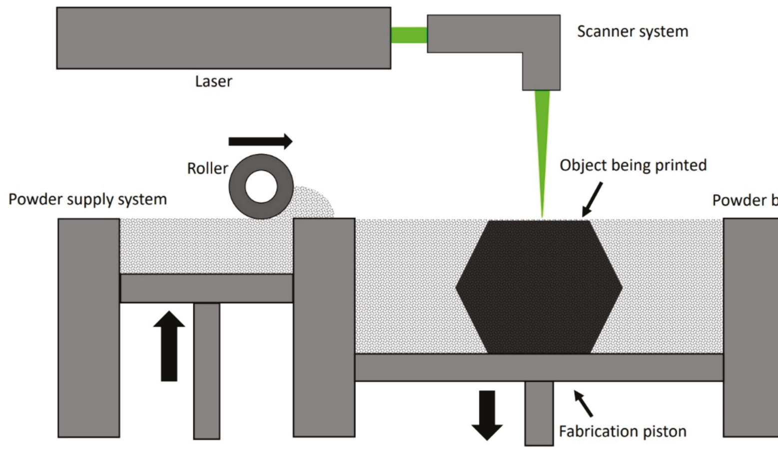

Figure 5. Scheme of SLM fabrication process.

Modern SLS can produce objects in layers with a thickness of 30–200 μm, depending on the particle size of the powder [49,50]. Once a structure is finished, the remaining powder must be brushed or blown off. Depending on the laser's power and the used material, the particles are sintered or fully melted together. In SLS, the emitted laser melts the grains' surface, while the core stays unaffected and solid. This partial melting allows for binding between the grains. Employing higher-powered lasers melts the grains fully to the core, resulting in homogenous structures, which solidify by cooling down [48]. Equipped with highly powered precision lasers, SLM improved many disadvantages of SLS, such as rough surfaces with high porosities and large shrink rates of the produced parts [48–50]. On the other hand, complete melting leads to higher surface tensions and therefore requires a smaller thickness of layers [48]. SLM-produced objects demonstrated high accuracy and precision in prosthodontic research, satisfying clinical requirements, sometimes surpassing other methods such as casting or milling [51–54].

Metal powders used for three-dimensional printing must fulfill all requirements regarding biocompatibility, thus allowing a safe mid- and long-term use in the oral environment. Therefore, alloys certified for oral application are commonly utilized for orthodontic purposes, among which, chromium-cobalt alloys [6,8,51,54,55] and titanium alloys for submucosal placements [48,50,55] are the most popular in orthodontics.

In the post-processing, the surface must be milled to remove all supporting structures as well as the rough outer oxide layer. Since SLS/SLM requires fewer supporting structures, post-processing is eased compared to other additive manufacturing procedures [49] (Figure 6A). If needed, preformed parts such as orthodontic screws, hooks for extraoral gears, or tubes can be added to the device by welding them to the framework (Figure 6B,C). Appliances designed with direct screws also require the matching thread to be welded onto the framework, because its co-axial construction is too complex to be printed.

Figure 6. Post-processing of a 3D-printed appliance. (**A**) Unprocessed printed parts, (**B**) surface milled and polished, Jack screw and tubes welded to the appliance, (**C**) completed appliance for molar distalization and simultaneous face mask wear (hooks).

Auxiliary devices such as insertion guides or models are usually manufactured using printing resin materials. Depending on the purpose and the chosen resin, printing technologies can be divided into light curing and fused deposition modeling (FDM) [6,55]. Photosensitive resins can be printed using specific light-curing techniques called stereolithography (SLA), digital light processing (DLP), and photo jet (PJ). In SLA and DLP, objects are shaped by applying selective UV-light irradiation to a reservoir of liquid resin. The polymerized resin is attached to a platform that moves at defined distances away from the light source, resulting in an incremental shaping of objects similar to the SLM process. In contrast, in PJ technologies, photopolymers are sprayed onto a platform by a horizontally moving print head. The light curing occurs simultaneously through a UV-light-emitting lamp positioned at the print head. Posteriorly, the platform moves away on the Z-axis in the desired amount of layer thickness, shaping the object in incremental layers. In light-curing 3D printers, almost any kind of liquid photopolymers can be used. By applying PJ, printing composites of resins and ceramics is possible. In FDM, objects are shaped by heating thermoplastic materials up to their melting points and printing filaments in layers. Besides conventional thermoplastic materials, biological thermoplasts, such as polylactic acid, polycaprolactone, or acrylonitrile-butadiene-styrene, can be employed in FDM [6,55]. If the printed object will be employed in the oral cavity and may come in direct contact with blood, the material must be sterilized prior to usage and requires the respective certification. Commonly used materials for printed medical devices can be sterilized using ethylene oxide, e-beam, hydrogen peroxide, gamma radiation, or steam sterilization in an autoclave, depending on the chosen polymer [56–61] and the national legal requirements. Moreover, a high degree of biocompatibility must be ensured for materials used in the oral cavity, regarding a lack of cytotoxic, sensitizing, or irritational properties.

4. Sources of Error

Printed appliances show very high precision, good fit, and low error rates. Nonetheless, digital workflows and clinical implementation are complex and require specific operational knowledge and exact working methods.

Depending on the chosen workflow and appliance installation mode, there are possible sources of error to be considered. In case that all steps have been accurately performed, guided insertion enables accurate insertion and appliance fits. However, minor errors may accumulate, and in this case implants may show a certain degree of variance from their planned position, resulting in poor congruence and thus requiring an adjustment of the device's connectors before installation. Position variance may occur due to the plastic guide's flexibility, if too much pressure forces the drills into a false angulation, or if the time period between the intraoral scan and the appointment of appliance insertion has been too long. In the latter case, the insertion template and the appliance itself can show variation due to a patient's natural growth or a change in the stage of eruption and tooth position. If the inconsistency between the implants and the connector is too big, tension will be placed on the connection, resulting in patient discomfort, implant overload, and potential implant failure. Nonetheless, the connectors can be minimally widened by using a milling cutter. Additionally, the rigid printing alloys also allow a certain degree of bending, leading to the possibility of readjusting and correcting minor inaccuracies. According to clinical experience, implants inserted into the patient's palate before the scan without a stabilizing appliance may trigger the patients' tongues to press against the implants, resulting in manipulation and a possible loss of the implants.

Direct Screw-borne appliances that are fixed onto the teeth prior to the implant insertion usually do not bear the risk of screw overloading or improper fit, as minor angular deviations can be compensated by the appliance. Nonetheless, insertion of the screws through the incorporated guides can generate moments and forces on the whole appliance when locking the screws into the thread of the connector, especially in case of angular discrepancies. This could stress implants and adhesive connections. As a result, the implants may loosen, or the adhesive bonds may weaken. Therefore, maintaining a correct insertion angle is strongly recommended. Similar to conventional appliances, printed metal devices can be subject to fracture or loss of adhesive connections to the teeth. Whereas screw-retained devices can be removed temporarily for repair, fracture of the direct screw-borne appliances may necessitate exchange of the implants whenever no intraoral repair is possible.

5. Clinical Application

5.1. Temporary Anchorage Device-Borne Appliances

When skeletal anchorage is employed, the non-digital workflow consists of unguided implant insertion, followed by a conventional silicone molding with impression caps. Then, the appliance is manufactured based on a plastered model with laboratory analogs, mainly consisting of preformed components. Once finished, the appliance can be installed unless it does not fit properly.

In contrast, for skeletally anchored appliances manufactured using a digital workflow, the clinician can choose and alter the mode of installation. The appliance can be designed to be fastened onto formerly placed implants, or it can be installed simultaneously with the implants using an insertion guide for the implantation, or it can be attached to the teeth prior to the implantation (*Direct Screws*). In the latter case, the appliance itself serves as an insertion template [33].

Taking a closer look into the various options of implant installation helps with outlining the respective advantages and disadvantages.

A common way to place an implant-supported appliance into the oral cavity is to set the implants at their determined position using an insertion guide (Figure 7A,B). This ensures a perfect match with the printed appliance (Figure 7C). Fixing the appliance to its skeletal anchor requires a mechanism that locks the appliance on top of the implants and prevents rotations. If a fixation screw is used, adapting the appliance, or changing to a completely different device, is possible [33] (Figure 7D). The adhesive bond between the appliance and teeth occurs before the insertion of the fastening screws.

In case of *Direct Screws*, the appliance itself serves as the guide [33] (Figure 8). In this case, the practitioner adhesively attaches the appliance to the teeth and then uses the hole of the appliance's connectors as an insertion aid. This installation method can be advantageous because it eradicates a possible mismatch between implants and the appliance, and eliminates the need for an additional 3D-printed insertion guide.

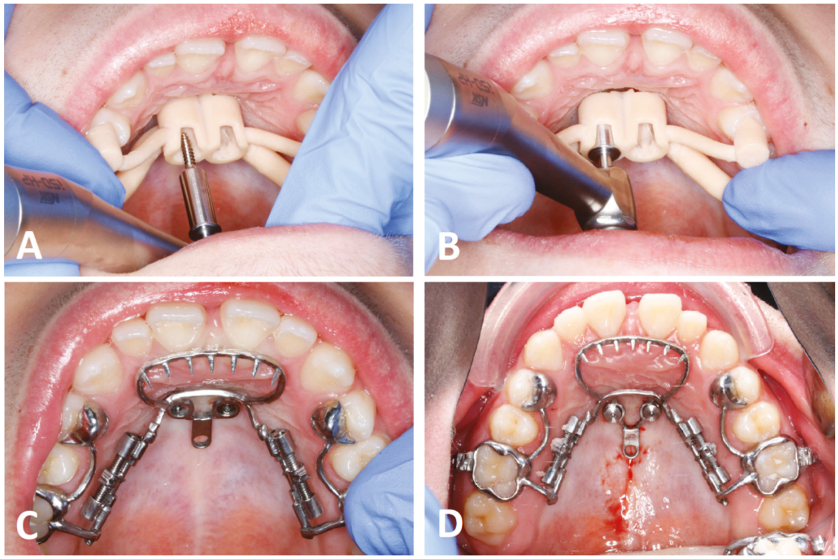

Figure 7. Appliance installation with guide: (**A**) Guided implantation. (**B**) Vertical control is achieved through a stop between the guide and the insertion tool. (**C**) Try-in on implants. (**D**) Appliance fastened on two implants by fixation screws.

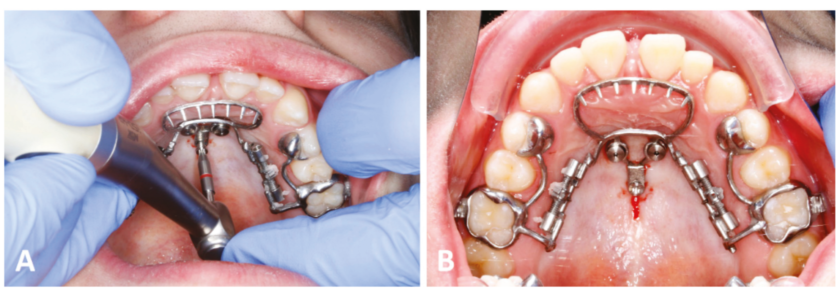

Figure 8. Direct screw insertion: (**A**) Implant placement with appliance itself serving as a guide. (**B**) Completed installation of the appliance from Figure 7 on two mini-implants (with fixation screws) and a posterior direct mini-implant inserted through the appliance itself.

When implant placement is performed prior to intraoral scanning, appliance installation is performed similar to conventional approaches. In case of immediate loading, one must consider that primary stability of the implants reduces within the first 4–6 weeks after insertion due to a loss of primary stability. Nonetheless, immediate loading of the implants has been reported to be not detrimental or even beneficial in terms of implant stability [62,63].

The molar coupling of digitally printed appliances can be achieved by means of printed shells or palatally attached tubes. These couplings do not penetrate the attached soft tissue and usually remain coronally to the interproximal contacts [64]. It may be speculated that this can lead to higher failure rates, which, however, can be reduced by etching the surface of the teeth and using an adhesive system instead of temporary cement. Additionally, no rubber-based separation of teeth is required prior to appliance insertion [7,65]. The preformed shape of customized, 3D-printed molar shells thus obviates the need to adapt bands, which may increase comfort for both the patient and practitioner [7,40]. Even though limited evidence is available, the digitally printed shells seem to provide clinically acceptable survival rates [7].

5.2. Computer-Aided Manufactured Mentoplates

Conventional *Mentoplates* were shown to be effective, especially in case of maxillary protraction in early class III treatments [28,29]. Using computer-aided manufacturing, *Mentoplates* can be designed to hold supra-constructions with various functional elements of high precision, including sliding mechanics for distalization or mesialization of posterior teeth, which increase the range of treatment options (Figure 9).

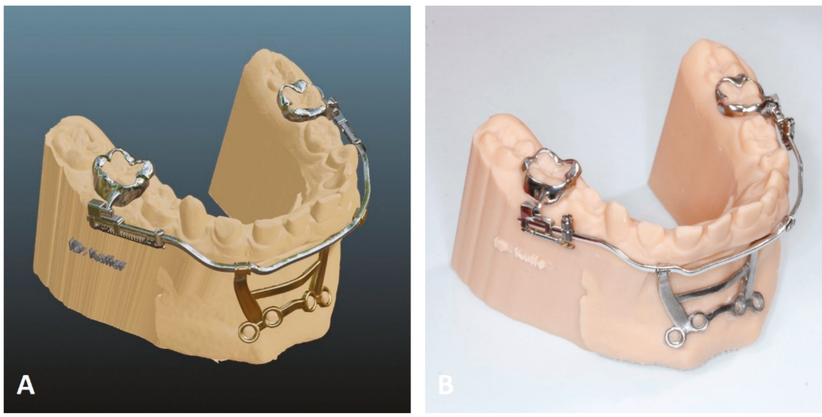

Figure 9. 3D-printed *Mentoplate* with mesial-slider supra-construction: (**A**) Digital design. (**B**) Finished printed appliance (software: Blender, Blender Foundation).

Both conventional and digitally manufactured appliances with a submucosal extent can be made from titanium alloys. Under local anesthesia or narcosis, the oral surgeon prepares a mucoperiosteal flap, revealing the bone structures of the chin. Conventional *Mentoplates* are adapted by bending them manually to fit the chin bone. The two extension arms are shortened and formed either straight or bent into a hook. Depending on the purpose of the treatment, they should penetrate the soft tissue at the mucogingival border or within the attached mucosa to avoid infection and eventually loss of the appliance.

3D-printed appliances do not require adaptation within surgery as they are already customized regarding the individual anatomical situation. While guides for palatal appliances lead the drill during the implants' insertion, osteosynthesis screws are directly applied into their connectors. Therefore, the guide must stabilize the *Mentoplate* in the correct position while the screws are inserted. For this purpose, it proved efficient to assemble the *Mentoplate* with its supra-construction and employ the posterior shells as orientation. To achieve stable triangular support, a printed guide can be mounted onto the lower incisors as well as on the supra-construction, ensuring vertical control (Figure 10). After fastening the screws, the guide can be removed, and the active elements can be adjusted as indicated. After the screws are applied, the flap can be adapted and sutured.

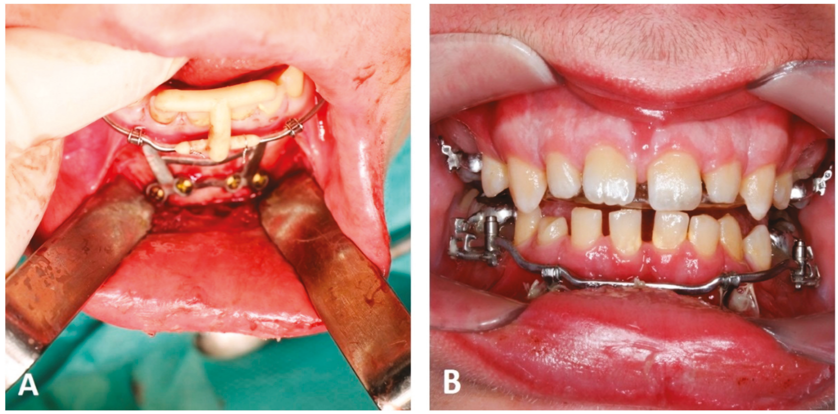

Figure 10. Clinical installation of a CAD/CAM manufactured *Mentoplate* with mesial-slider supraconstruction. (**A**) Surgical insertion and anterior vertical control with guide. (**B**) Intraoral situation with installed appliance.

6. Conclusions

The incorporation of a digital workflow with computer-aided design enabled fabrication of customized 3D-printed metal orthodontic appliances. In complex cases requiring additional anchorage, the incorporation of skeletal anchorage may increase the spectrum of treatment options and enable highly time-efficient treatments with increased patient comfort. Nonetheless, digital workflows necessitate an initial training period and may be associated with high acquisition costs, especially when applied in-house. As errors during the process can accumulate from the different steps, accurate planning and validation of a proper fit prior to insertion of screws and appliances is strongly recommended.

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manually conducted by a trained technician, aligners may not be perfectly homogenous even if they are fabricated on the same cast [20].

Few techniques exist to assess homogeneity and microstructure of thermoplastic materials. Micro-computed tomography (micro-CT), whose laboratory usage has been introduced in the 1990s for structural analysis of calcified tissues, provides high resolution three-dimensional images from various specimen [21]. Its applicability to assess aligner material properties has been demonstrated recently [22–25].

Therefore, the present study aimed at assessing the impact of dental model height on PET-G aligner thickness values, and at investigating the within-group variability of thickness values potentially arising from the manual fabrication process.

2. Materials and Methods

2.1. Study Design

The present study reports on a sample of n = 20 polyethylene terephthalate glycol (PET-G) aligners. The aligners were thermoformed on a 3D-printed dental model of the upper jaw with either narrow (N) base height, or high (H) base height (achieved by placing a spacer of 5 mm height) (n = 10 aligners per group, respectively). The perpendicular distance between occlusal plane and model-tray were defined as model height and amounted to 11 mm (Figure 1). Consequently, model height in the N group was 11 mm, and 16 mm in the H group.

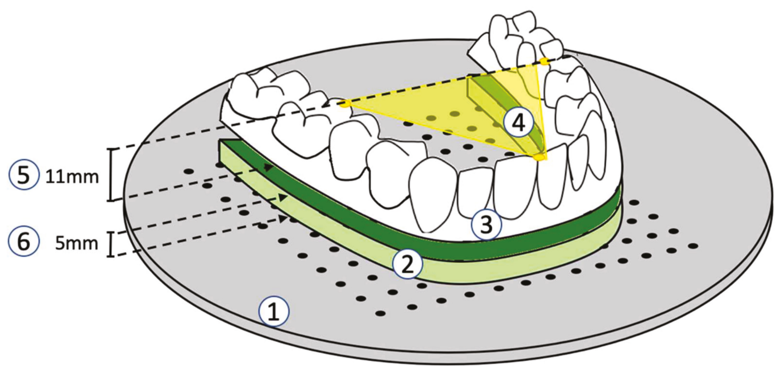

Figure 1. Thermoforming-Setup (1: model tray; 2: spacer; 3: acrylic dental arch model; 4: occlusal plane parallel to the model tray; 5: model height = 11 mm; 6: spacer height = 5 mm).

The 3d-printed model was fabricated using a SLA printer (Form 2, Formlabs Inc., Somerville, MA, USA) at 100 micron resolution (100% infill) and using the Draft Resin (Formlabs Inc.). The model was oriented in a nearly vertical position to minimize warping. The model was manually cleaned and cured according to manufacturer's recommendations. After curing, the model base was manually grinded to compensate any warping and ensure full flat contact to the thermoforming tray. It was then stored at ambient air at 20 °C and 50% humidity.

2.2. Thermoforming

Thermoforming was achieved using a Biostar thermoforming machine (Biostar VII, Scheu Dental, Iserlohn, Germany) and a PET-G raw material of 0.5 mm thickness (CA Clear Aligner, Scheu Dental, Iserlohn, Germany). The dental model was placed in the center of the thermoforming chamber such that the mid palatal suture was located in 12 o'clock position. The occlusal plane was oriented parallel to a perforated custom model-tray (Figure 1). The dental model was held in place by 3 positioning pins to ensure constant localization and orientation for each thermoforming process.

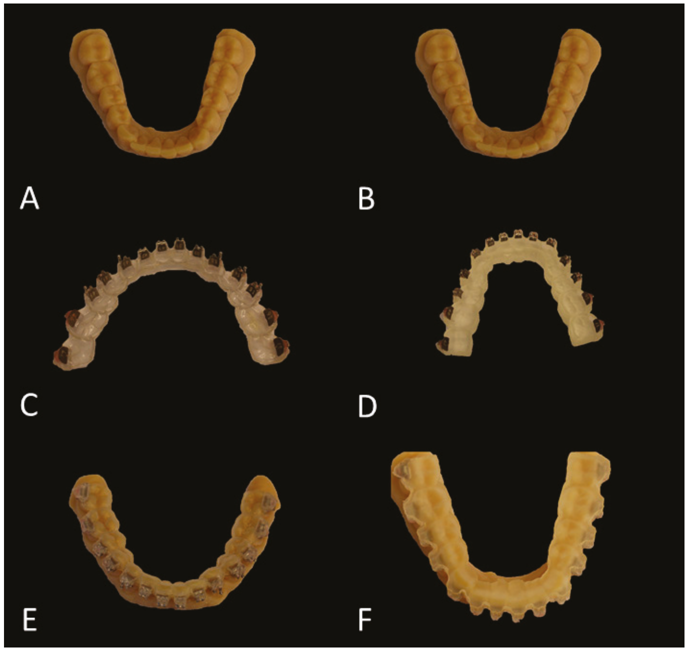

Figure 2. Overview of the bracket placement workflow. (**A**,**B**) Printed study models (SM). (**C**,**D**) Hard (**C**) and soft (**D**) tray with brackets placed inside the respective molds. (**E**,**F**) Printed casts with the brackets indirectly bonded by means of the hard (**E**) and soft (**F**) transfer tray.

The SM were imported into a proprietary software (OnyxCeph^{3TM} Lab software, Image Instruments, Chemnitz, Germany) for virtual placement of brackets/tubes utilizing the FA-Bonding module as follows: teeth 35–45 (Discovery smart bracket, Dentaurum, Ispringen, Germany), teeth 36, 37, 46, 47 (Ortho-Cast M series buccal tubes, Dentaurum, Ispringen, Germany).

The FA-Bonding module operates semiautomatically and utilizes the facial axis (FA) for automated bracket positioning along the vertical tooth axis. Minor adjustments were performed by one single experienced orthodontist (RJ). The SM together with the virtually placed brackets served as reference models (SMref) (Figure 3).

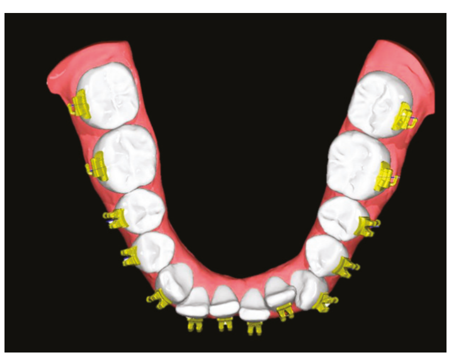

Figure 3. Reference model consisting of a digitized study model and the semiautomatically positioned brackets/tubes (SMref).

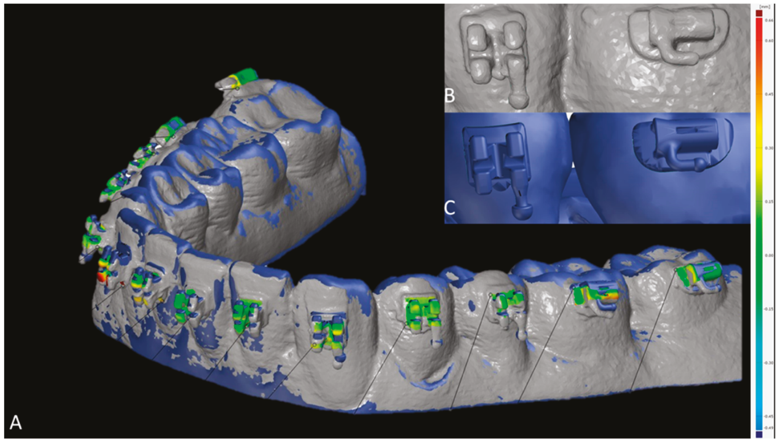

Figure 4. Measurement of the bracket transfer accuracy with GOM software. (**A**) Superimposed study models scanned with intraoral scanner (SMtest-IO) visualized in grey and reference study models (SMref) visualized in blue; distances are coded by heatmap coloring. (**B**) Scanned brackets from SMtest-IO in higher magnification. (**C**) Original brackets from SMref in higher magnification.

2.5.2. Micro-CT Scanning (Method Validation)

Digitization of SMtest was also achieved using a micro-CT (Viva CT 80, Scanco Medical, Brüttisellen, Switzerland). The scans were performed at 70 kVp, 114 µA, 535 ms integration time and 2× frame averaging, and reconstructed to a nominal isotropic voxel size of 39 µm. The so achieved SMtest-mCT models were segmented and surfaces were extracted using Amira software (v2019, Thermo Fisher Scientific, Berlin, Germany).

SMtest-mCT surfaces were aligned with the respective SMref models using a landmark-based registration procedure followed by an iterative closest point algorithm (Meshlab software) as described earlier [26]. Owing to metal artefacts on the micro-CT scans, each scanned bracket on the SMtest was replaced with the respective original bracket surface by superimposing the latter on the micro-scanned ones (Amira software). This technique enabled preservation of the true positions of the brackets on the digitized SMtest models (Figure 5A,B).

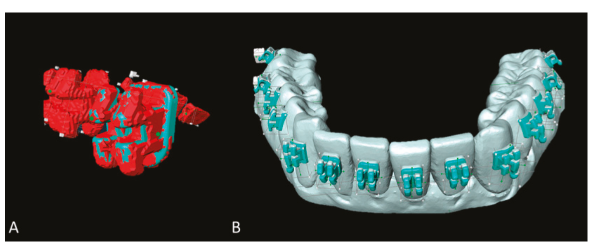

Figure 5. Superimposition and image post-processing with Amira software. (**A**) Micro-CT scanned brackets (red) and the superimposed original brackets (green). (**B**) Reference study models (SMref) visualized in grey and corrected brackets from micro-CT scanned study models (SMtest-mCT) visualized in turquoise.

Bracket placement accuracy on SMtest-mCT models was assessed using Fusion 360 (Autodesk, San Rafael, CA, USA). All brackets and tubes were virtually separated from SMref and oriented along the global Cartesian coordinate system, and an additional local coordinate system was defined for every bracket/tube (Figure 6A). The disagreement between brackets from SMtest-mCT and SMref along the x-, y- and z-axes was recorded as

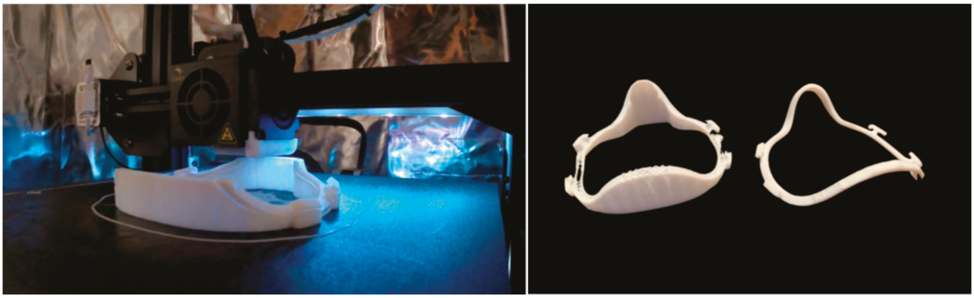

Figure 4. A 3D-printing process with a personal filament 3D-printer (left) and the Mask Fitter after printing and finishing.

Figure 5. The Mask Fitter adaptation to the scanned face anatomy.

For the present study, a total of 20 dental staff were recruited for this service evaluation selected following "convenience sample" selection criteria, and they voluntarily agreed to try the Mask Fitter. A written consent was obtained from each of the participants, by which they agreed to participate in the study and to have their data used as part of the research. They were face-scanned and then their Mask Fitters were 3D-printed as previously described. The emailed .stl files were processed for slicing with the software Ultimaker Cura 4.7.1 (Ultimaker BV, Utrecht, The Netherlands) and 3D-printed with the printer Anycubic Mega S (Anycubic, Shenzhen, Guangdong, China) set with Generic Polylactic Acid (PLA), Profiles 0.1 mm and Infill 100%. PLA and Acrylonitrile Butadiene Styrene (ABS) are the recommended filament, but there are no specific restrictions providing the printed surfaces are not porous. PLA and ABS can be disinfected by immersion in the disinfecting solution recommended for each material. The Mask Fitter is placed on the external side of the surgical mask, and thereby does not need any special biocompatibility or antiallergic properties. For the present survey, the filament 1.75 mm PLA 3D-Printer Filament (Anycubic, Shenzhen, Guangdong, China) in grey shade, was used. This is a biodegradable filament made from lactic acid, produced via starch fermentation during corn wet milling. The available information for this material is: Diameter 1.75 $\pm$ 0.02 mm, Tensile strength $\geq$55 MPa, Hardness HRC 105–110, Density 1.25 g $\pm$ cm^3. For the present study, it has been 3D-printed with the nozzle temperature at 210 °C and printing platform at 60°.

design was saved as the reference scan STL file (RS-STL), which was then used to fabricate 3D-printed (n = 30) and milled (control) (n = 10) molar crowns.

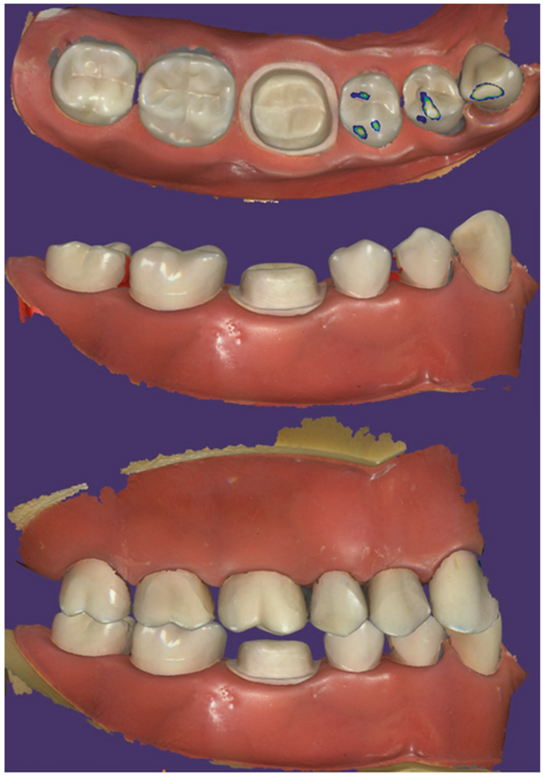

Figure 1. Mandibular right first molar preparation.

2.2. Crown Fabrication

Three different layer thicknesses (20 µm, 50 µm, and 100 µm) were used to print the crowns (n = 10 per layer thickness). First, the RS-STL file of the crown was imported to DLP software (MoonRay S100, SprintRay Inc, Los Angeles, CA, USA) to arrange the build angle and support configuration. As recommended by the manufacturer of the printing resin material (Nextdent Crown and Bridge Micro Filled Hybrid-MFH, C&B; 3D systems, Soesterberg, The Netherlands), the occlusal surface of the crown was angled 45° from the print area for improved occlusal surface details. The semiautomatically created support structures were checked and supports that were automatically created on the margin area and fitting surfaces of the crowns were manually eliminated. Then, this configuration was duplicated 10 times and 10 identical crown configurations were arranged in the build platform of the DLP printer (MoonRay S100 Software, SprintRay Inc, Los Angeles, CA, USA) to print all crowns in the same configuration. This configuration was further saved for 3 different layer thicknesses to print the crowns in identical configuration, but by using different layer thicknesses. The crowns were printed with the DLP printer (MoonRay S100, SprintRay Inc, Los Angeles, CA, USA) and an interim printing resin material (N1 shade, Nextdent Crown and Bridge Micro Filled Hybrid-MFH, C&B; 3D systems, Soesterberg, The Netherlands, Lot: XH312N21) by using 20 µm, 50 µm, or 100 µm (n = 10) layer thickness. According to the manufacturer, the printing material has 100–130 MPa flexural strength, 2400–2600 MPa flexural modulus, ≤70 µg/mm^3 sorption, and ≤15.5 µg/mm^3 solubility.

The DLP printer has UV DLP projector, LED-based light source, and 405 nm blue-violet light resin curing unit. After printing, the printed crowns were removed from the platform by using a putty knife, ultrasonically rinsed for 5 min (first 3 min then 2 min) in 96% clean alcohol solution (Alcohol isopropilico, Quimi Klean, Mexico City, Mexico) according to the manufacturers' recommendations. After ensuring that the crowns were dry and free of alcohol residue, they were postpolymerized by using an ultraviolet polymerizing unit (SprintRay Procure Model SRP1811A, SprintRay Inc, Los Angeles, CA, USA) (405 nm LED arrays) for 30 min (Figure 2).

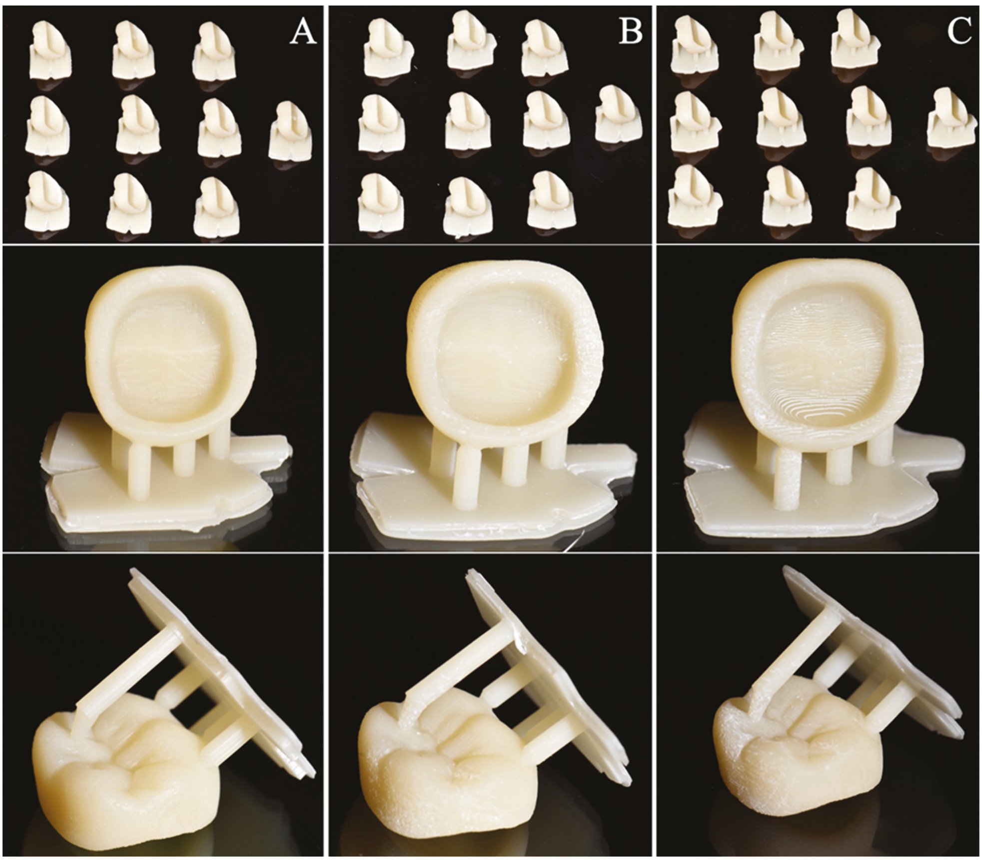

Figure 2. 3D-printed interim crowns ((**A**): 20 µm crowns; (**B**): 50 µm crowns; (**C**): 100 µm crowns).

The support structures were cut and trimmed after cooling and the surface was gently smoothened to prevent errors during the alignment procedure.

In the milling technique, crowns were milled (Wieland Zenotec mini, V6.12.04, Wieland Dental + Technik GmbH & Co.KG, Pforzheim, Germany) from a polymethyl methacrylate (PMMA) block (A2 shade, Lot number: HL201104, Upcera, Shenzhen Upcera Dental Technology Co. Ltd., Shenzen, Guandong, China). The designed STL file was inserted in the block to mill 10 identical crowns. The support structures were cut and trimmed after milling, and the support surfaces were gently smoothened. All crown fabrication processes were performed by one operator (G. Ç).