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Additive Manufacturing: A Novel Method for Fabricating Cobalt-Chromium Removable Partial Denture Frameworks

Keywords

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ABSTRACT

Additive manufacturing (AM) often referred to as 3D printing (3DP) has shown promise of being significantly viable in the construction of cobalt-chromium removable partial denture (RPD) frameworks. The current paper seeks to discuss AM technologies (photopolymerization processes and selective laser melting) and review their scope. The review also discusses the clinical relevance of cobalt-chromium RPD frameworks. All relevant publications in English over the last 10 years, when the first 3D-printed RPD framework was reported, are examined. The review notes that AM offers significant benefits in terms of speed of the manufacturing processes however cost and other aspects of current technologies remain a hindrance.

INTRODUCTION

Removable partial dentures (RPDs) are an affordable and popular alternative to fixed restorations such as implants and fixed partial dentures (FPD)/bridges.[1,2] Their use dates back to around 700 BCE when human or animal teeth were anchored to gold bands. This was long before ivory and vulcanite dentures were used.[3, 4] Currently, RPD frameworks are constructed mainly with cobalt-chromium (Co-Cr) alloys due to their low cost and durability compared to gold-based alloys.[5] The construction of these frameworks is however time-consuming and requires exceptional manual dexterity to minimise potential errors.[6] These limitations are claimed to be less problematic with significant improvements in additive manufacturing (AM) technologies.[7, 8] AM which is also known as three-dimensional printing (3DP), is defined as 'a process of joining materials to make objects from 3D model data, usually layer upon layer, as opposed to subtractive manufacturing methodologies'.[9] In essence, 3D printing can simply be described as two-dimensional printing of materials in successive layers until the desired 3D part is completed.

Eggbeer *et al.*[6] are documented as the first to build resin RPD frameworks with a stereolithography apparatus (SLA-250) machine which was popular at that time. These frameworks were then cast in a cobalt-chromium (Co-Cr) alloy using conventional methods. The first Co-Cr RPD framework manufactured directly by selective laser melting (SLM) for a patient is credited to Williams *et al.*[10] For this review, the term selective laser melting refers to all laser-based metal AM systems utilizing 'full melting' mechanisms to build Co-Cr parts not specific to any particular variant. Despite the evident advantages of AM, using computer aided design and computer

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aided manufacturing (CAD/CAM) systems requires adequate operator training, as well as prior knowledge of traditional manufacturing methods. Furthermore, acquiring CAD/CAM systems requires additional expense, which could be a hindrance to smaller dental laboratories especially when manufactured parts still require considerable manual finishing. In light of this, the sustainability of these technologies depends on how they compare with traditional 'lost-wax' or investment casting methods. Their acceptance as a suitable alternative depends on accessibility, affordability and clinical evidence.[11] The current paper seeks to discuss the AM technologies used in removable partial prosthodontics, and review their scope. All relevant publications in English over the last 10 years when the first 3D-printed RPD framework was reported are examined.

ADDITIVE MANUFACTURING TECHNOLOGIES

Digital manufacturing by AM or 3DP involves feeding CAD model files (usually STL files) into a designated 3D printer to build 3D parts. For most applications, it is worthwhile to consider the print speed, part cost, ease of use and build resolution of the 3D printer.[12] For the sake of conciseness, only significant aspects of photopolymerization processes for castable resin/wax frameworks and selective laser melting techniques for direct metal frameworks are discussed in this review. The complete manufacturing process is discussed in great detail in relevant literature and manufacturer's reference sources.

PHOTOPOLYMERIZATION PROCESSES

In photopolymerization, liquid photopolymer resins undergo a chemical reaction upon irradiation, usually in the ultraviolet (UV) range to become solid.[13] Currently, stereolithography (SL), digital light processing (DLP) and material jetting are the three photopolymerization processes that could be used to manufacture resin/wax RPD frameworks for investment casting.[14]

Stereolithography (SL) is considered the pioneer of AM processes.[15,16] In a typical SL system, layers of liquid photopolymer resin from a vat are selectively cured or solidified layer-by-layer with an ultraviolet (UV) laser beam[17] to form resin frameworks. Parameters that influence the performance and functionality of SL parts in 3DP include physical and chemical properties of the photopolymer resins, speed and resolution of the optical scanning systems, the power, wavelength and types of laser used, the spot size of the laser, the recoating system and the post-curing process.[18]

Digital Light Processing (DLP) is similar to SL but their parts are built in a shallow vat of resin resulting in less waste and lower running costs. This process also uses a more conventional light source, such as an arc lamp, with a liquid crystal display panel or a deformable mirror device, which is applied to the entire surface of the vat of photopolymer resin in a single pass, generally making it faster than SL.[19] Resin frameworks built with these technologies require support structures in their liquid vats which are removed after post-curing. Post-curing involves placing the manufactured frameworks in a solvent solution to remove any wet resin remnants, followed by final curing in a UV oven which fully hardens these frameworks and allows them to maintain their structural integrity.[17]

Material jetting is a process whereby a liquid photopolymer is selectively squirted through multiple jet heads, and then cured with a passing of UV light as each layer is deposited. This process is noted for producing accurate frameworks with a very smooth finish.[19] Unlike SL and DLP, no post-curing is required in material jetting. Some 3D printers have the capacity to produce a single RPD framework from multiple materials which have different characteristics and properties. Support structures are melted away in a heated oven[20] in some processes while others require cautious removal of the support structures with a water jet or by hand to avoid breakage or distortion.[21] A 3D-printed resin framework and cast Co-Cr framework are shown in Figure 1.

Figure 1: 3D-printed resin and cast Co-Cr RPD frameworks. Source: 3D Systems

Although RPD frameworks built by photopolymerization require further processing, this represents a cost-effective option for dental laboratories that cannot afford higher-priced SLM machines. It is possible in some instances to invest multiple frameworks in a casting ring to reduce time and material cost.[22] However, with substantial investment casting steps required, using this manufacturing technique will primarily be dependent on its economic viability and other advantages it may offer a particular laboratory.

SELECTIVE LASER MELTING

Selective laser melting (SLM) is a powder bed fusion process that uses high-energy laser beams to melt metallic powders together, in a layer-by-layer fashion, into 3D objects.[8] The laser beam is traced across a powder bed of tightly compacted powdered material, according to the 3D data fed into the machine, in the X and Y axes. As the laser interacts with the surface of the powdered material it fuses the particles to each other thereby forming a solid mass. After a layer of metallic part is built, the powder bed drops incrementally (in Z-axis) and a roller levels the powder over the surface. New layers are built on top of the existing layer successively until the process is completed in an enclosed gas chamber.[19,23,24] The different SLM technologies[25-27] are able to produce almost 100% component density.[13] In dental technology, the SLM process eliminates many physical manufacturing steps such as spruing, investing, burnout, casting and devesting. SLM as an advanced manufacturing process also offers reduced labour and speed compared to AM/investment casting technique. It is claimed to be suitable for constructing a standard RPD framework in a time period of around 30 minutes.[25]

The ease of processing titanium alloys, which previously required specialised equipment in conventional casting methods[27,28] is an advantage. However, high cost associated with these metal 3D printers (e.g. EOS M270 is estimated to cost $560000), makes them less financially viable[14] for smaller laboratories, which may be compelled to outsource these procedures to third-parties. The stages of manufacturing the Co-Cr removable partial denture by the SLM process are shown in Figure 2. Framework finishing and denture processing are accomplished in the same manner as standard conventionally manufactured frameworks.

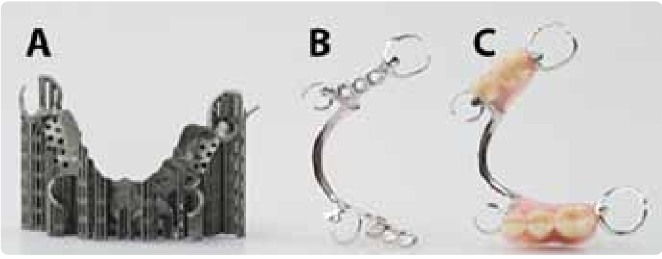

Figure 2: RPD Co-Cr framework with support structures after SLM B. Support structures removed and finishing accomplished in the same manner as standard conventionally manufactured Co-Cr frameworks C. Finished Co-Cr RPD framework. Source: EOS GmbH

CLINICAL PROPERTIES OF SLM-PRODUCED CO-CR ALLOYS

Documented studies show SLM-produced Co-Cr alloys possess adequate mechanical properties.[29-31] for clinical use and are superior to cast Co-Cr alloys.[32-34] These mechanical properties are closely related to their microstructure. Yager *et al*[35] found SLM-produced Co-Cr alloy possess superior homogeneity with a smaller grain size (~5–80µm) compared to cast Co-Cr alloy (200–300µm). The small grain sizes in the SLM process increases the yield strength of the alloy[3] and is formed as a result of rapid solidification and little segregation of alloying elements.[36] This differs from investment casting, where within the pool of molten ingots, larger grains are formed around tiny nuclei that grow until the grain boundaries meet in the solid state. As a result of this, dendritic structures may be very large in cast Co-Cr alloys, where the size of a single grain can approach the diameter of a removable partial denture framework clasp.[3] A comparision of tensile test properties between cast and SLM-produced Co-Cr alloys in a study by Alifui-Segbaya[32] is shown in Table 1.

Overall, superior mechanical properties of dental alloys translate into higher resistance to distortion and efficient transmission of occlusal forces to the remaining teeth or other tissues.[5] These properties are also important to accomplish a tissue-friendly satin finish which is achieved via electrolytic polishing to prevent fitting and cleaning problems on the fitting surface of Co-Cr frameworks.[37] A good surface finish in alloys serves as an important corrosion resistance parameter.[7,8] Corrosion is a prerequisite for biologic effects to take place in the human body.[38,39] In the assessment of biocompatibility, corrosion data is a useful adjunct to cytotoxicity studies.[7] Corrosion tests showed that SLM-produced Co-Cr alloys have low ion emission rates[7,40,41] under ISO 10271 acceptable threshold of 200µgcm-² within 7±0.1 days.[42] Cytotoxicity tests also proved to be safe, non-irritant and nontoxic to oral tissues and the body as a whole.[43,44] Figure 3 shows corrosion trends of SLM and cast Co-Cr alloy samples after 42 days. In this study,[32] the SLM-produced Co-Cr alloys performed better than cast Co-Cr alloys. Interestingly, the electrobrightened samples in both groups also performed better in the corrosive environment than hand-polished samples. The two prepared finishes were carried out to simulate the clinical use of Co-Cr frameworks.

Acceptable clinical fit of denture frameworks is achievable with SLM techniques[10,45,46] but there is inconclusive evidence of this. Kim *et al*.[47] recorded inferior fits for SLM-produced dental bridges compared to cast dental bridges. It must be noted that fitting accuracy of dental devices can also be influenced by other pertinent clinical and laboratory processing steps[29] as well as the build parameters of SLM systems.[18]

Differences in the build parameters have been identified as the likely cause of the differences in the microstructure of SLM-produced Co-Cr alloys.[33-35] Yager *et al*.[35] found that at low energy densities, small grains without preferred orientations form, while at high energy densities, long columnar grains form with strong texture along the travel direction of the laser due to a large thermal gradient that appears as the laser moves across the sample. Also, smaller cellular dendrites in SLM-produced Co-Cr alloys have been linked to higher cooling rates in SLM systems that use fiber lasers compared to CO^2 lasers.[33] The metallurgical conditions under which these alloys are processed differ substantially from those in conventional casting hence it is possible that the different microstructures can subsequently alter the physico-mechanical properties of individual components that make up SLM-produced frameworks.[35] In light of this, further microstructure studies on SLM-produced Co-Cr are required as the current knowledge is lacking in conclusive evidence.

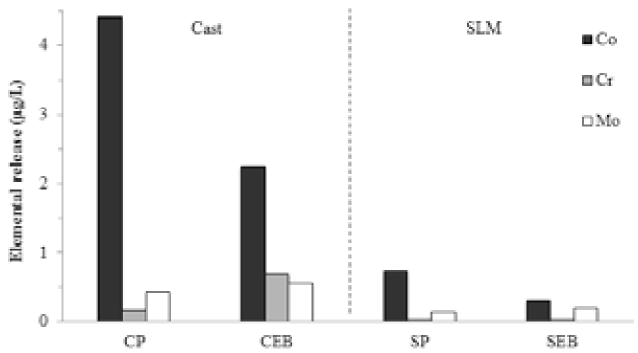

Figure 3: Elemental release of Co-Cr-Mo in artificial saliva after 42 days in accordance with ISO 10271. Key: CP = cast and polished Co-Cr; CEB = cast and electrobrightened Co-Cr; SP = SLM-produced and polished Co-Cr; SEB = SLM-produced and electrobrightened Co-Cr. Source: Alifui-Segbaya (2011)

Table 1. Tensile test properties of cast and SLM-produced Co-Cr alloys. Source: Alifui-Segbaya (2011)

Cast Co-Cr	Max.	Min.	Mean	Median	CV (%)	SD
Max Load (N)	6663.8	6093.2	6438.7	6421.1	3.1	200.5
Deflection at Max. Load (mm)	2.9	1.2	1.8	1.9	31.1	0.6
Work to Max. Load (J)	13.6	3.9	7.4	6.3	45.6	3.4
Stiffness (Pa)	12348000	11771000	12011000	12040000	1.7	209520
Load at Break (N)	666.4	609.3	643.8	642.1	3.1	20.1
Deflection at Break (mm)	2.9	1.3	1.9	2.1	29.6	0.6
Work to Break (J)	13.9	4.3	7.8	7.1	43.1	3.4
SLM-produced Co-Cr	**Max.**	**Min.**	**Mean**	**Median**	**CV (%)**	**SD**
Max Load (N)	9677.6	95550.3	9585.9	9570.9	0.4	42.4
Deflection at Max. Load (mm)	3.3	2.6	2.9	2.8	8.9	0.3
Work to Max. Load (J)	22.6	18.5	19.7	19.3	7.2	1.4
Stiffness (Pa)	14498000	11962000	13353000	13635000	7.4	983290
Load at Break (N)	967.7	955.1	958.6	957.09	0.4	4.3
Deflection at Break (mm)	3.4	2.7	3.1	2.9	8.7	0.3
Work to Break (J)	23.3	19.3	20.3	19.9	6.9	1.4

CONCLUSION

This review has shown that:

SLM is a manufacturing process which offers increased speed and flexibility by eliminating many physical steps as needed in investment casting. Thus, a decision to choose a particular technique over another may depend on its economical viability. Although both techniques require manual finishing, SLM-produced frameworks offer better predictable clinical outcomes as this process is digitally controlled and thus offers a standardized method for the manufacture of dental devices. These devices are then likely to be much closer to the manufacturer's specification, than those produced by investment casting which as a process is fraught with many variations. As 3D printing gains in popularity, it is imperative for practitioners to be familiar with this technology. Educational institutions could integrate 3D printing modules into their curricula to enable newly graduated dental technologists and clinicians to fully understand the range of AM technologies available to them. This will allow them to make informed decisions regarding the most appropriate techniques, systems and materials to choose from for varying dental applications.

CONFLICT OF INTEREST STATEMENT

We affirm that we have no financial affiliation (e.g., employment, direct payment, stock holdings, retainers, consultantships, patent licensing arrangements or honoraria), or any involvement with any commercial organization with direct financial interest in the subject or materials discussed in this manuscript, nor have any such arrangements existed in the past three years. Any other potential conflict of interest is disclosed.

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