Effect of Different Finish Line Designs on the Marginal and Internal Fit of Metal Copings Made by Selective Laser Melting Technology

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EFFECT OF DIFFERENT FINISH LINE DESIGNS ON THE MARGINAL AND INTERNAL FIT OF METAL COPINGS MADE BY SELECTIVE LASER MELTING TECHNOLOGY

by

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ABSTRACT

EFFECT OF DIFFERENT FINISH LINE DESIGNS ON THE MARGINAL AND INTERNAL FIT OF METAL COPINGS MADE BY SELECTIVE LASER MELTING TECHNOLOGY

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Introduction: Marginal fit has been defined as the gap between the prepared tooth and the intaglio surface of the restoration. Internal gap is the perpendicular measurement from the internal surface of the casting to the axial wall of the preparation. Selective laser melting has been used for fabrication of metal copings such as Co-Cr base alloys and Au-Pt noble alloys. The purpose of this study was to determine the effect of different finish line designs on the marginal and internal fit of metal copings made from high noble, 25% noble and base alloys manufactured by SLM technology.

Material and Methods: An ivorine right maxillary central incisor was prepared with three different finish line designs. Three preparations were scanned using a Trios scanner and a total of 90 dies were printed using DPR 10 Resin. Ninety metal copings were fabricated using 3 different types of alloys. Copings were cemented to the dies using resin cement. All specimens were sectioned buccolingually using a low speed diamond saw. Marginal and internal gaps were measured at 5 locations. Marginal and internal gap images were determined using an inverted bright field metallurgical microscope at x 100 magnification. A two-way multivariate analysis of variance (MANOVA) was conducted to determine overall significance followed by analysis of variance (ANOVA) for each dependent variable ($\alpha=0.05$).

Results: Overall, 2700 measurements were obtained for the study. The result of statistical analyses indicated that both alloy type and finish line had a significance influence on overall fit of the copings. For the internal fit, the alloy type had a significant effect ($p<0.001$), but the finish line had no statistically significant influence($p=0.337$). For the marginal fit, both the alloy type and the finish line had a statistically significant effect, ($p<0.001$). There was no statistically significant interaction between variables.

Conclusions: Finish line types did not significantly influence the internal fit between the copings and the dies, whereas alloy type did influence the fit between copings and dies. SLM-fabricated copings made with the Base Alloy (Co-Cr) on teeth prepared with deep chamfer finish lines demonstrated the best marginal fits when compared to the other groups.

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CHAPTER I

INTRODUCTION

Marginal and Internal Fit

One of the factors affecting longevity of fixed dental prostheses (FDP) is dependent upon accurate fit of the prosthesis.(1) An important clinical assessment for success of a FDP is the marginal fit of the crown or retainer.(2–4) Marginal fit has been defined as the gap between the prepared tooth and the intaglio surface of the restoration.(5) The marginal fit can also be described as the linear distance between the finish line of the preparation and the margin of the restoration.(6) Holmes et al. defined the internal gap as the perpendicular measurement from the internal surface of the casting to the axial wall of the preparation.(7) Marginal misfit of the prosthesis could eventually lead to failure of the prosthesis.(3) A large marginal gap will lead to the use of excess luting agent and upon exposure to the oral environment, it may decompose due to moisture and chemomechanical processes.(8) As a result, microleakage may lead to secondary caries, and if the tooth is vital it could lead to pulpal inflammation or necrosis.(3,4,8–10) Inadequate adaptation of the crown margins may lead to more plaque retention, subsequent subgingival microflora which may lead to gingival and periodontal issues.(11) Another consequence of marginal misfit would be a decrease in the strength of the restoration due to stress concentrations.(12)

Some authors have discussed clinically acceptable marginal gaps. In a 5-year clinical study where 1000 metal-ceramic crowns were examined, McLean and Fraunhofer concluded that a marginal gap no greater than 120 µm was clinically acceptable; Christensen conducted a linear regression prediction formula and concluded 39 μm was the least acceptable marginal discrepancy.(13–15) Other authors have written that marginal discrepancies between 100 and 150 μm are clinically acceptable.(13–15)

In-depth studies regarding marginal and internal fit of restorations fabricated using computer aided design/computer aided manufacturing (CAD/CAM) systems have been performed. An in vitro study by Bindl and Mormann evaluated the fit of crown copings prepared by 4 different CAD/CAM systems (CEREC inLab, DCS, Decim, and Procera); it was demonstrated that the marginal gaps ranged between 17 to 43 μm and internal gaps ranged between 110 to 136 μm.(1) Another in vitro study was conducted by Hyun-Soon et al., where the marginal gaps for zirconium oxide based crowns fabricated by Digident and Lava CAD/CAM systems was evaluated. It was reported that mean marginal gaps ranged between 82 to 83 μm.(16) Reich S et al. examined the marginal and internal fit of 3 unit FDPs fabricated using Digident, Vita In-Ceram, and Lava CAD/CAM systems. It was found that the marginal gaps ranged from 67 to 92 μm and internal gaps ranged from 105 to 383 μm.(17)

Reich S. et al. performed a study on single crowns made by a chairside CAD/CAM system; the results yielded mean marginal gaps of 100 μm and internal gaps that ranged from 148 to 284 μm.(18) Marginal gaps of single cast crowns has also been studied; 50 % of the marginal gaps of the studied crowns exceeded 150 μm.(19)

Another study by Quante K et al. reported marginal and internal fit of metal-ceramic crowns fabricated with a laser melting procedure (BEGO Medical, Bremen, Germany). Their results resulted in mean marginal gap widths ranged from 74 to 99 μm.(20)

Luting Cements

Dental luting agents or cements forms the link between a restoration and the tooth structure.(21) Although it is of high importance to establish retention and resistance forms during tooth preparation, dental cement may be used to act as a barrier against microbial leakage by sealing the interface between tooth and restoration and holding them together through some form of surface attachment.(22) This attachment could be mechanical, chemical or both. An ideal dental adhesive should possess favorable compressive and tensile strength, have sufficient fracture toughness to prevent dislodgment, exhibit adequate film thickness and viscosity to ensure complete seating, be tissue compatible, demonstrate good working and setting time, and provide a durable bond between dissimilar materials.(23–25)

In 1878, Pierce invented zinc phosphate cement, which is considered the oldest dental luting agent. It has the longest track record as a luting agent for securing cast restorations. For more than 130 years, it has served as a standard by which newer systems are compared to.(26,27) In 1903, silicate cements were developed. They were the earliest tooth colored restorative materials. Silicate cements could be considered to be the precursors to modern composite resin and glass ionomer cements.(27)

Polyacrylate cement were discovered in 1968 by D.C. Smith, where he used zinc oxide as a powder and polycarboxylic acid as the liquid component. It was the first cement system to be developed with the potential for adhesion to tooth structure.(28) In an attempt to combine both properties of silicate and polycarbxylate cements, Wilson and Kent developed glass ionomer cement in 1969.(28) Then came resin modified glass ionomer cement, which were developed in 1986.(27)

In the mid 1980's, resin cement was invented. Resin cements with dentin bonding agents have shown greater retention of restoration to teeth when compared to zinc phosphate cement.(28) Resin cements can be classified according to their method of polymerization, and can be classified into auto-polymerizing, dual-polymerizing and light-polymerizing cements. Auto-polymerizing cements are recommended for use in areas difficult to reach with light curing units such as metal restorations.(29) Dual-polymerized cements are polymerized by both a chemical reactions and visible light of specific wavelengths. Dual-polymerized cements contain a self-initiator (benzoyl peroxide) and a light initiator (camphoroquinone).(30) Lastly, light polymerized cements are cements that set only with exposure to certain wavelengths of visible light. They contain a photo-initiator similar to camphoroquinone although some cements may contain different types of photo-initiators.(30)

According to the American Dental Association (ADA) specification No. 8, luting cement film thickness for a single crown restoration should not exceed 25 µm when using a Type I luting agent, and should not exceed 40 µm when using a Type II luting agent.(31) Type I luting materials are designed for the accurate seating of precision restorations such as inlays. Type I luting agents include hydroxyapatite, glass ionomer, zinc phosphate, and polycarboxylate cements. Type II luting materials are designed for all uses except for cementing precision restorations and require increased film thicknesses.(32)

Die Spacer

In the past, dentists and researchers believed that having a frictional fit between the coping and the tooth surface would achieve more retention. This meant that during cementation, a perfect fit couldn't be obtained due to lack of space for the luting agent.(33–36) Die spacers are designed to allow space for the cement between the internal surface of the restorations and the tooth surfaces. This space reduced the stress areas created during cementation and allowed for a better fit and retention for definitive restorations.(36)

In 1993, Grajower et al. stated that "an optimum fit of the casting can be obtained only if relief space allows for the cement film thickness and roughness of the tooth and casting surfaces". They believed that an effective technique included placing a spacer directly to the die, including the base of the tapered region. They recommended that the only part not to be included was the horizontal part of the shoulder finish line. They also arbitrarily recommended that 50 µm be used as the thickness of die spacers.(37)

Tjan and Li found that an improved marginal fit was achieved when resin cement was used when compared to the marginal fit obtained with zinc phosphate cement. They speculated that the reason could be because, in their study, they applied two layers of copal varnish to the surfaces of the prepared teeth prior to cementation with zinc phosphate cement, which could have influenced the marginal fitting of the metal castings.(38) In a study reported by Anna Olivera et al. showed that resin cement (Panavia 21) exhibited the highest tensile strength when compared to resin modified glass ionomer cement (Vitremer luting cement) and zinc phosphate cement (Harvard Richter and Hoffmann, Berlin Germany).(39) These results were also in agreement with the results obtained by Lee and Swartz, Tjan and Li, Pamieijer and Jefferies, El-Mowafy et al. and Gorodovsky and Zidan.(38,40–43)

<u>Lost Wax Technique</u>

Lost wax casting is an ancient technique for replicating an object by casting it in molten metal. The lost wax technique has been used in dentistry for more than 100 years and is still one of the most popular methods for fabricating metal dental restorations.(44) This is a process where a wax pattern of a dental restoration is made and converted to a casting alloy or a ceramic.(45) Many alloys have been designed for use in dentistry; Cobalt/chromium (Co-Cr) alloys can be cast similar to nickel/chromium (Ni-Cr) alloys and have better corrosion resistance.(46,47) Metal structures are conventionally fabricated using lost-wax technique. However, CAD/CAM technology allows the precise design of metal structures.(48)

<u>Dental Alloys</u>

For successful cast restorations, alloys should meet minimum requirements for strength, stability, castability, corrosion/tarnish resistance, burnishability, polishability and biocompatibility. Metal ceramic alloys must possess additional physical properties above and beyond the properties of non-metal ceramic alloys. Success of metal ceramic restorations is dependent upon the physical properties of the metal substructures.(49) These alloys require higher melting temperatures, thermal compatibility with ceramics, oxide formation and sag resistance.(49) According to the ADA in 1986 dental cast alloys are divided into different groups:(50)

1. High noble alloys ≥ 60 % Au, Pt, Pd and ≥40 % Au
2. Noble Alloys ≥ 25 % Au, Pt, Pd
3. Base metal alloys < 25 % Au

Noble-metal metal-ceramic alloys (Gold-Platinum-Palladium):

Gold-platinum-palladium (Au-Pt-Pd) alloys were the first alloys successfully used for metal-ceramic restorations; however due to high costs, more economical alloys were developed with significantly better mechanical properties and sag resistance. If the alloy had more palladium than platinum, it was referred to as a gold-palladium-platinum alloy (Au-Pa-Pt). When palladium was eliminated from the alloy, the alloy would be referred to as a gold-platinum alloy (Au-Pt).(51) Because of their properties having a low sag resistance, those alloys should be limited to single crowns and three unit FDPs.(52) Tables 1 and 2 below list the properties of several noble-metal and metal-ceramic alloys.

Table 1. Compositional ranges (wt. %) of noble-metal metal-ceramic alloys.

Type	Au	Pt	Pd	Ag	Cu	Sn	Ga	In	Other
Au–Pt–Pd	75–88	≤8	≤11	≤5	–	2–5	–	<1	Fe, Re
Au–Pd	44–55	–	35–45	–	–	8–12	≤5	8–12	Ru, Re
Au–Pd–Ag	39–77	–	25–35	12–22	–	3–7	–	1.5	Fe, Ru, Re
Pd–Ag	–	–	50–60	28–40	–	4–8	–	1–5	Ru
Pd–Cu	≤2	≤1	70–80	–	9–15	0–8	3–9	0–8	Ru
Pd–Ga	0–2	–	74–85	1–7	–	—	6–10	6	Ru

* Adapted from Powers and Sakaguchi.(53)

Table 2. Properties of noble-metal metal-ceramic alloys.

Type	Ultimate tensile strength (MPa)	0.2% yield strength (MPa)	Elastic modulus (GPa)	Elongation (%)	Diamond pyramid hardness (kg/mm2)	Casting temperature (◦C)
Au–Pt–Pd	480–500	400–420	81–96	3–10	175–180	1150
Au–Pd	700–730	550–575	100–117	8–16	210–230	1320–1330
Au–Pd–Ag	650–680	475–525	100–113	8–18	210–230	1320–1350
Pd–Ag	550–730	400–525	95–117	10–14	185–235	1310–1350
Pd–Cu	550–1100	550–1100	94–97	8–15	350–400	1170–1190

* Adapted from Powers and Sakaguchi.(53)

Gold Palladium Silver (Au-Pd-Ag) Alloys:

Au-Pd-Ag alloys were developed to overcome several limitations associated with Au-Pt-Pd alloys, including high cost, low hardness, and poor sag resistance.(51) These alloys can be subdivided into 2 main groups; high silver and low silver. An alloy is considered a high silver containing alloy when it contains 12 % silver (Ag) or more and it is considered a low silver containing alloy when it contains 5 % to 11.9 % silver (Ag).(52) The major drawback of silver-containing alloy is the potential for silver to discolor the porcelain.(52,54)

Gold-Palladium (Au-Pd) Alloys:

These alloys were developed to minimize limitations associated with silver and the high coefficient of thermal expansion of Gold-Palladium-Silver (Au-Pd-Ag).(51) Coefficient of thermal expansion is defined as the change in length per unit of the original length of a material when its temperature is raised 1° K.(52) In 1977, these alloys generally exhibited a white gold color and were commercially successful.(52,54) The main limitation of Au-Pd alloys was an incompatible degree of thermal expansion with some high expansion porcelains. Due to this limitation, multiple Au-Pd alloys were developed that contained less than 5 % silver. Castability of these alloys improved, thermal expansion increased, as well as their clinical usefulness.(54)

Palladium Cobalt (Pd-Co) Alloys

These alloys had limited clinical usefulness. The main benefits associated with Pd-Co alloys included high coefficients of thermal expansion which made them compatible with certain types of dental porcelains.(54) Manufacturers have added 1-2 percent of noble metals such as gold and/or platinum to improve its grain structure. The major limitation associated with Pd-Co alloys was the tendency to form a dark oxide layer which tended to discolor the porcelain. It was also reported that these alloys had weaker bonding with porcelain than did Pd-Cu alloys.(55)

Base-Metal Ceramic Alloys

There are two main categories of this type of alloy: nickel-based and cobalt-based, (Tables 3 and 4). Alloys in both categories contain chromium as the second largest metal in the alloy; chromium is involved with improved corrosion resistance.(51) Base-metal alloys have excellent physical properties. For example, they exhibit the highest modulus of any alloy type used for cast restorations.(56) The modulus of elasticity is defined as the measure of the stiffness or rigidity of an alloy, since it corresponds to the amount of stress for unit elastic strain.(52)

Table 3. Compositional ranges (wt. %) of base-metal metal-ceramic alloys.

Type	Ni	Cr	Co	Ti	Mo	Al	V	Fe	Be	Ga	Mn	Nb	W	B	Ru
Ni–Cr	62–77	11–22	–	–	4–14	0–4	–	0–1	0–2	0–2	0–1	–	–	–	–
Co–Cr	–	25–34	53–68	–	0–4	0–2	–	0–1	–	0–3	–	0–3	0–5	0–1	0–6

*Adapted from Powers and Sakaguchi.(53)

Table 4 (Properties of base-metal metal-ceramic alloys).

Type	Ultimate tensile strength (MPa)	0.2% yield strength (MPa)	Elastic modulus (GPa)	Elongation (%)	Diamond pyramid hardness (kg/mm2)	Casting temperature (°C)
Ni–Cr	400–1000	255–730	150–210	8–20	210–380	1300–1450
Co–Cr	520–820	460–640	145–220	6–15	330–465	1350–1450

*Adapted from Powers and Sakaguchi.(53)

Base-metal alloys used in metal-ceramic restorations, have exhibited better castability than noble alloys. (55) However, they have a tendency to form thicker, darker oxide layers than do noble metal alloys, which may present esthetic challenges.(56) Historically, base-metal alloys were divided into 4 groups: nickel-chromium-beryllium, nickel-chromium, nickel-high-chromium, and cobalt-chromium.(56)

1. Nickel-chromium-beryllium alloys were used due to the presence of beryllium which facilitated casting.(55) This type of alloy has been discontinued due to health concerns.
2. The major contents of Ni-Cr alloys are nickel and chromium, they may also contain minor amounts of other metals.(51) Commercially available Ni-Cr alloys are close in composition and physical properties but differ in corrosion resistance.(56) Aluminum and titanium have been added in small amounts to form strengthening precipitates. Iron, tungsten and vanadium have also been added for solid solution hardening. Of the elements added for hardening these alloys, molybdenum and tungsten are the most effective.(57)
3. Cobalt is the main component in cobalt-chromium (Co-Cr) alloys. Chromium has been added for strength and corrosion resistance.(51) Co-Cr has been established as a satisfactory alternative for patients known to be allergic to nickel.(56) Co-Cr alloys have the highest melting range of the casting alloys. This limitation makes it a little difficult to manipulate while casting in the laboratory.(56)

Intra-oral Scanners

In 1987, the first commercially available digital intraoral impression system was invented, it was known as CEREC 1 system.(58) Its method of operation was based on the principle of "triangulation of light", and the surface being scanned required a coat of powder to improve the scan quality.(59) After that, multiple new digital intraoral devices were developed. CEREC, Lava^{TM} C.O.S, iTero, E4D and

TRIOS are some of the available intraoral digital impression systems available in the market today.(60)

Dentist' experiences and patient compliance are a key factors in the quality of the digital impressions.(61) Multiple studies have evaluated the clinical behavior of FDPs fabricated using the intraoral digital impressions and CAD/CAM protocols. These studies have demonstrated acceptable qualities in the restorations including marginal fit and occlusion characteristics.(62)

Intraoral digital impressions have improved over time and are now able to record complete arches. Intraoral digital scanners allow the dentist to record/capture teeth, implant scan bodies, and soft tissues in 3 dimensions. CAD/CAM has changed the way dentistry is practiced and has become an integral part of dental practice.(63,64)

Clinicians seeking to overcome the shortcomings associated with conventional elastomeric impressions have used digital impressions as an alternative to elastomeric impression materials and procedures. One major advantage of digital impressions is having the ability to magnify the impression digitally, highlight the defective areas in real time, and recapture missing areas.(65)

Intraoral cameras work either by recording images in a video type format or by recording still images during the scanning process. Still photos are based upon triangulation or parallel confocal laser scanning. Lava C.O.S (3M ESPE) and Lava True Definition scanner (3M ESPE) uses active wavefront sampling for data collection from which a video image is formed. CEREC AC Bluecam (Sirona) uses active triangulation and optical microscopy to produce still images. The CEREC AC Omnicam (Sirona) uses video for data collection. iTero and 3Shape Trios uses the parallel confocal method to produce digital images.(66)

CAD/CAM

Over 40 years ago, CAD/CAM processes were introduced for several dental applications, and included designing and milling ceramic inlays and veneers.(67) Since the development and evolution of CAD/CAM technology at the beginning of the 1970's, the accuracy of dental restorations made using this technology has increased and the cost per unit has decreased as the cost of the milling machines decreased. (44)

There are many CAD/CAM systems available for processing different types of dental restorations in dental clinics, dental laboratories and manufacturing centers.(1,68,69) Three pioneers contributed to the development of CAD/CAM systems in dentistry.(68) In 1971, Dr. Duret has been identified as the first pioneer in dental CAD/CAM and began fabricating crowns by incorporating the shape of occlusal surfaces using a series of systems that began with an optical impression of the abutment tooth made intra-orally. This was followed by designing an optimal crown form taking into consideration functional movements, and milling the crowns using a numerically controlled milling machines.(68,70)

A second pioneer, and developer of the CEREC system, was Dr. Werner Mormann. His technology was utilized chair-side directly on patients. Following tooth preparation, he directly captured (imaged) the preparations using an intra-oral camera. An inlay could be designed and milled from a ceramic block in a compact in-office milling machine. Due to the capability for one-day fabrication of CAD/CAM restorations, CAD/CAM technology rapidly spread throughout the profession and dental laboratory industry. (68,71)

In the early 1980's, Dr. Andersson developed the Procera system. His development began as a method that used cobalt chromium alloys as a substitute for gold alloys. This change dramatically decreased costs. Many people are known to be allergic to certain metals, especially in northern Europe. Dr. Andersson researched using titanium as a substitute for cobalt chromium alloys. Due to difficulties associated with casting titanium, he attempted to fabricate titanium copings using spark erosion existing technology and introduced CAD/CAM technology into the process of composite veneered restorations.(72) "This was the application of CAD/CAM in a specialized procedure as part of a total processing system. This system later developed as a processing center networked with satellite digitizers around the world for the fabrication of all-ceramic frameworks. Such networked production systems are currently being introduced by a number of companies worldwide."(68,73)

Subtractive vs. Additive Manufacturing

Most of the fabrication techniques in CAD/CAM technology have been based upon subtractive manufacturing, or in another word, milling technology.(44) It is an approach where the material is removed to create a desired shape, the desired shape is created effectively but at the expense of materials discarded as wastes during the process. This is a major limitation associated with milling technology as waste material adds to the cost of fabrication of restorations.(74) Additive manufacturing processes have been recently introduced. This provides a completely new concept, "it was developed to meet the requirements of rapid manufacturing (RM) and rapid prototyping (RP), such as stereolithography (SLA), fused deposition modeling (FDM), selective electron beam melting (SEBM) or selective laser sintering (SLS)".(47–51) Each of those techniques have been used for fabrication of restorations using different dental materials.

SLS has been increasingly used for fabrication of dental restorations.(44) SLS is basically a process that fabricates 3-dimensional (3D) parts by incorporating layers of powders of different materials (such as polymers, ceramics or metals), under the heat of a focused laser beam. The process is driven by the data provided by the CAD file.(76,77) Terminology has not yet been clearly identified in the dental field, but according to the binding mechanism of the sintered material, researchers have preferred to use the term SLS for non-metallic materials such as ceramics or polymers, others have used the term DMLS (direct metal laser sintering) or SLM (selective laser melting) for alloys.(44,76,77)

Selective laser melting first started in the aerospace and automotive industries for fabrication of sophisticated hollow structures. This process was later modified and implemented in the dental field.(44) SLM is an additive manufacturing procedure, which manufactures metal parts directly from a 3D CAD model. Koutsoukis et al. stated “it works by fusing fine layers of metal powders by means of a high-power source of a focused laser beam. The concept of this technique is similar to that for SLA, except that in SLM the liquid medium has been replaced by the metallic powder.”(75)

The principle that SLM systems operate upon is that a 3D file of the desired object (dental restoration), created by a CAD system, is divided into vertical or horizontal layers and then transferred to the laser sintering device. The desired alloy powder is applied to form the platform, while the laser scanner scans the required surfaces according to the information gathered from the 3D CAD file. A powerful CO_2 laser is usually used because it can generate enough heat to sinter the powder and form a layer of metal. “The build platform is driven by a piston with the ability to adjust to the vertical axis. Adjacent to the manufacturing piston is the powder-feeding piston, capable of vertical adjustment. When operating, the laser beam transfers heat to the powder mixture, resulting in local melting and fusing of the particles. When the layer with the desired shape has been completed, the manufacturing piston backpedals while the feeding piston rises to refill the build platform, assisted by a roller. The procedure is then repeated for the next layer, until the product has been completely fabricated as designed by the 3D CAD file."(75)

Depending upon the properties of the alloy to be used for sintering, the parameters such as melting temperature, laser beam absorption/reflection coefficient and thermal conductivity should be noted. The average grain diameter of the powder could affect the mechanical properties of the restoration and metallurgical phenomena during solidification.(77,80) In order to minimize porosities and improve the mechanical properties, full melting of the powder particles is required.(77) Settings of the apparatus such as the scanning speed, the holding time, the temperature of the preheated bed and the thickness of each layer will all affect the quality of the final result.(68–70) One important aspect in the SLM process is minimizing potential thermal distortion, which could be accomplished by improving wettability based on proper selection of the preheated bed temperature.(76)

Takaichi et al. studied the microstructure of SLM surfaces and they compared it to castings and milled surfaces. They reported that there was a significant difference between the surfaces of SLM, milled and cast Co-Cr alloys. It was concluded that cast Co-Cr alloys have the characteristic dendritic microstructure with a dispersed heavier phase in interdendritic positions, while the milling microstructure depends solely on the characteristics of the block used and SLM surfaces are dependent mainly on operational parameters.(80)

Porosities are undesirable when it comes to fabricating dental restorations as it causes the deterioration of the mechanical properties of the metal.(83) SLM and milling techniques are superior to castings when it comes to porosities. In theory, SLM technique could provide structures with up to 100 % nominal density of the sintered alloy but it depends mainly on the proper adjustment of operating conditions including laser per, scan spacing, scan rate and scan thickness.(40, 36, 22) Porosities in the castings on the other hand could be due to shrinkage of the castings, and the gross dendritic structure of Co-Cr alloys during solidification.(83,85,86) Porosity in milled structures is mainly dependent upon the initial quality of the metallic block.(87)

Selective laser melting (SLM) has been used for fabrication of metal copings such as Co-Cr base alloys and Au-Pt noble alloys.(20,88) One of the first SLM systems was accurate to approximately 50 to 80 µm per layer thickness.(89) Progressive development of the SLM process has led to better results. Multiple studies reported layer thicknesses of approximately 20 µm for dental applications.(60–63)

Preparation Finish Lines

Clinically, the effect of different finish line designs on fitting accuracy should be taken into account and should be meticulously studied.(93) Several studies examined the effect of different finish lines on adaptation of crowns and yielded contradictory results.(94) For cast restorations, Preston and Schillingburg recommended beveled shoulders as the best type of finish line for cast restorations.(95,96) For In-Ceram crowns, Pera et al. reported that chamfer or 50-degree shoulder tooth preparations yielded better marginal adaptation when compared with 90-degree shoulder finish lines.(97) Comlekoglu et al. compared the marginal gaps associated with zirconia crowns designed with knife edge, mini chamfer, chamfer and rounded shoulder finish line designs and found that the lowest marginal discrepancy values was for knife edge finish line (87 µm) compared to mini-chamfer (114 µm), chamfer (144 µm) and rounded shoulder (114 µm) finish line designs.(98) Euan at al. found a lower mean marginal gap value for Lava all-ceramic system crowns designed with round shoulder finish lines compared to chamfer finish lines.(99) On the contrary, Tsitrou et al. found that there was no significant difference in marginal gaps of dental restorations designed with shoulder and chamfer finish lines.(100)

For Procera crowns, Lin et al. reported that featheredge finish lines resulted in increased marginal discrepancies when compared with 0.8 mm rounded shoulder and 0.5 mm rounded shoulder finish lines.(101,102) In another study by Gwinner FP et al., it was reported that crowns fabricated with sintered gold copings, beveled long chamfer (BLC) finish lines showed less marginal gaps when compared to beveled round shoulder finish lines (BRS).(103) Ates et al. concluded in their study that cast Co-Cr crowns had the best adaptation on chamfer finish lines whereas CAD-CAM Y-TZP frame works had the best adaptation on shoulder finish lines.(104)

CHAPTER II

AIM OF THE STUDY

Multiple studies have examined the marginal fit of copings formed using SLM technology. However, none of the studies used standardized finish lines to test the fit of the copings. Moreover, several of the studies were performed with the copings fitted on chamfer finish lines, while others placed the copings on heavy chamfer finish lines. Additionally, none of the studies demonstrated which material was best to be used when SLM technology is utilized.(20,88,91,105,106)

The purpose of this study was to determine the effect of different finish line designs on the marginal and internal fit of metal copings made from high noble, 25% noble and base alloys manufactured by SLM technology.

Two null hypotheses were considered for this study: (1) finish line design will have no effect upon marginal accuracy or internal fit of SLM restorations; and (2) composition of the metal alloy will have no effect upon marginal accuracy or internal fit of SLM restorations.

CHAPTER III

MATERIALS AND METHODS

Preparation:

An ivorine right maxillary central incisor (T1560; Columbia Dentoform Corp) was prepared to receive a metal coping, Fig. 1.

Figure 1. (Unprepared right maxillary central incisor.)

Three different finish line designs were prepared using diamond burs (Brasseler USA):

1. Shoulder with a 90 degrees axiogingival internal line angle (S)
2. Deep Chamfer (DC)
3. Chamfer (C)
    - Preparations were standardized with incisal reduction of 2 mm for all three groups.
    - Uniform axial reduction of 1.5 mm for groups (S) and (DC), and 1 mm for (C).

- Margin width was 1 mm for groups (S) and (DC) and 0.5 mm for (C).
- Total convergence angle was 12 degrees for all groups

An index was made of the unprepared teeth using polyvinyl siloxane impression material (Express putty; 3M ESPE, St Paul, MN, USA) to standardize and measure the preparations, Fig. 2.

Figure 2. (PVS putty index of unprepared right maxillary central incisor.)

Measurements were made using a calibrated manual periodontal probe (UNC 15, Hu-Friedy, Chicago, USA), Figs. 3 and 4.

Figure 3. (Preparation index placed onto a prepared right maxillary central incisor.)

Figure 4. (UNC 15, Hu-Friedy Probe.)

12 degree total convergence was maintained by placing the typodont on a cast holder and placing all the parts on a surveyor. The 12 degree convergence was maintained by using the bur (Brasseler, USA) and the survey arm of the surveyor (J.M Ney Co., Bloomfield, Conn.), Fig. 5.

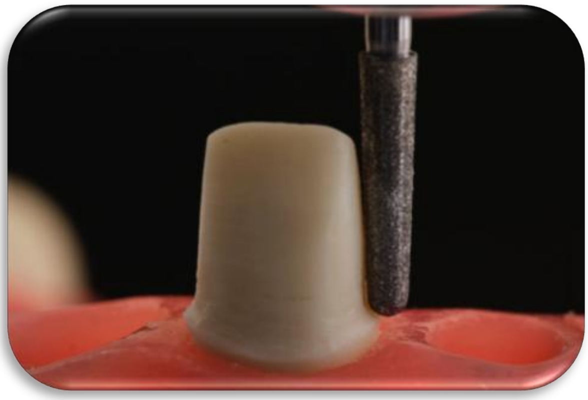

Figure 5. (Bur placed in the survey arm and held perpendicular to the long axis of the tooth.)

Before duplication of the prepared ivorine teeth, each tooth was attached to a square base fabricated using orthodontic resin (Dentsply Intl). This material increased the diameter of the ivorine tooth shaft and aided in mounting the tooth during sectioning, Fig. 6.

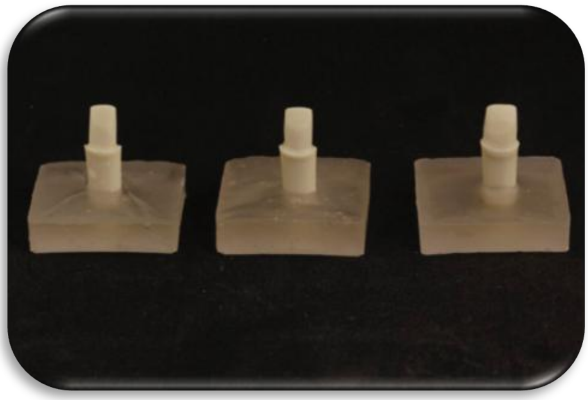

Figure 6. (Prepared teeth mounted in orthodontic resin base.)

Die Fabrication

The three preparations, along with the mounted bases, were scanned using a Trios scanner (3Shape, Copenhagen, Denmark) to create a stereolithographic (STL) file. All models were printed using DPR 10 Resin (Carbon3D, USA), Fig. 7.

Figure 7. (Printed die.)

Copings Fabrication

3Shape CAD design system was used to locate the margins and design the copings. Die spacer thickness of 25 µm was assigned uniformly to all the copings.

SLM Technology

Group B dies were manufactured from a base alloy; there were 10 specimens per tooth preparation. Group H dies were manufactured from a high noble alloy; there were 10 specimens per tooth preparation. Group N dies were manufactured from a 25% noble alloy; there were 10 specimens per tooth preparation. There was a total of 90 teeth in the study.

SLM metal copings were printed using a CAD/CAM system by Argen, (Argen Manufacturing System; Argen Corporation). Ninety metal copings were fabricated using 3 types of alloys; 30 copings were made from a base alloy-(Argen Manufacturing System; Argen Corporation) (Group B); (Co 61, Cr 25, Mo 6, W 5, Si <1, Fe <1, Mn <1), 30 copings were made from a high noble alloy, (Argen Manufacturing System; Argen Corporation) (Group H); (Au 40, Pd 39.9, Ag 10, Ru <1, In 10) and 30 copings were made from 25% noble alloy-(Argen Manufacturing System; Argen Corporation) (Group N); (Pd 25, Co 42.75, B <1, Mo 12, Cr 20).

Following fabrication, fit of the copings were checked visually with a light microscope at a magnification of 12.5× (Stereo Star Zoom, American Optical, Buffalo, NY). Internal adjustments were made as necessary to fit the master die, Fig. 8.

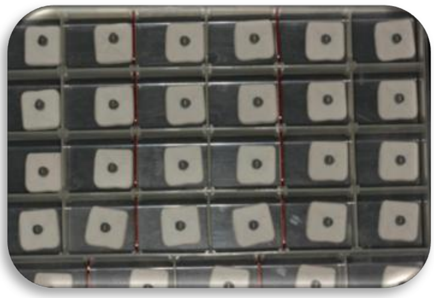

Figure 8. (SLM copings and dies.)

Table 5. (Alloys and finish line groupings.)

Groups	**(B)** - Base Alloy (N = 30)	**(H)** – High Noble Alloy (N = 30)	**(N)** – 25 % Noble Alloy (N = 30)
Finish Line Design	**(S)** - Shoulder with a 90 degrees axiogingival internal line angle **N = 10**	**(S)** - Shoulder with a 90 degrees axiogingival internal line angle **N = 10**	**(S)** - Shoulder with a 90 degrees axiogingival internal line angle **N = 10**
	(DC) – Deep Chamfer **N = 10**	**(DC)** – Deep Chamfer **N = 10**	**(DC)** – Deep Chamfer **N = 10**
	(C) – Chamfer **N = 10**	**(C)** – Chamfer **N = 10**	**(C)** – Chamfer **N = 10**
Total Samples	**N= 90**		

Copings Cementation

Resin Cement (Panavia 21 EX; Kuraray Noritake Dental Inc., Japan) was mixed according to the manufacturer's instructions, Fig. 9.

Figure 9. (Panavia 21 EX.)

After application of cement, the copings were seated with a rocking motion until they were completely seated on the die visually. The cemented coping-die assemblies were placed under an apparatus capable of maintaining a static deadweight load of 49 N; excess cement was removed using a fine microbrush prior to setting, Fig. 10.

Figure 10. (Cemented coping-die assembly placed under static load of 49 N.)

The cemented coping-die assemblies were kept under load for 3 minutes, as this was the setting time for the cement as per manufacturer instructions. After that, the specimens were placed into an incubator (Isotemp Incubator 655D, Fisher Scientific, USA) and was kept at 37 deg. C for 3 minutes to mimic mouth temperature and to ensure complete setting of each cement mix, Fig. 11.

Figure 11. (Isotemp incubator).

All specimens were then stored at room temperature until sectioning.

<u>Sectioning of Samples</u>

Each specimen was sectioned in a buccolingual direction using a low speed diamond saw (IsoMet speed saw; Buehler Ltd, USA) with a 127 × 0.4 mm diamond wafering blade (Buehler IsoMet, USA) under wet conditions, Fig. 12.

Figure 12. (Low-speed diamond saw with diamond wafering blade.)

After sectioning, each specimen was marked with a small notch using a small .010 mm round carbide bur (Brasseler USA) at points B, C and D in order to assist with orientation under high magnification, Fig. 13.

Figure 13 (Sectioned Tooth, with notches made at points B, C and D)

Measuring the Marginal and Internal Gaps

The marginal and internal gaps between the printed copings and each die for each sectioned specimen were measured at 5 locations, Fig. 14.

(A) Facial margin (Marginal Gap)

(B) Facial mid-axial (Internal Gap)

(C) Incisal (Internal Gap)

(D) Lingual mid-axial (Internal Gap)

(E) Lingual margin (Marginal Gap)

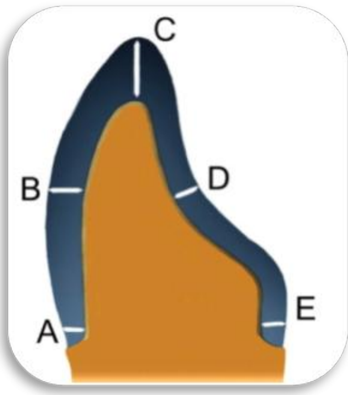

Figure 14 .(Tooth diagram of measured locations.)

Three measurements were made per point, for determining an average value at each point, which will total 15 measurements (3 × 5) per each half coping-die assembly. Two coping-die assemblies were produced from each specimen, Fig. 15.

Figure 15. (Specimen sectioned into halves.)

The marginal and internal gap images were determined using an inverted bright field metallurgical microscope at ×100 magnification (Metallograph/Microscope; Leco/Olympus), Fig. 16.

Figure 16. (Metallograph/ Microscope.)

The software used to calculate the marginal and internal gaps after the images were captured by the microscope was Spot Software 5.2 (Spot Imaging Solutions).

CHAPTER IV

STATISTICAL ANALYSIS

Means, standard deviations under different conditions were compared to test the null hypotheses. Box's test and Levene's test were performed to verify an assumption of equal variances. Material and type of finish line were used as independent variables and internal and marginal gaps were used as dependent variables. A two-way multivariate analysis of variance (MANOVA) was conducted to determine overall significance followed by analysis of variance (ANOVA) for each dependent variable (α=0.05). Tukey's HSD was used for post-hoc comparison (α=0.05). All statistical analyses were performed in SPSS (SPSS statistics 24, IBM).

CHAPTER V

RESULTS

As shown in Table 6, the mean of the Internal Fit of the Deep Chamfer finish line in the Base Alloy group showed the largest internal gap when compared with the other two finish lines. In the Noble Alloy group, chamfer finish lines had the largest internal gap when compared to the other two finish lines. In the High Noble Alloy group, the internal gap was largest with the deep chamfer finish line when compared to the other two finish lines.

As for the Marginal Fit, the mean measurement of the Base Alloy group showed the largest gap with Chamfer Finish Lines than the deep chamfer or shoulder finish lines. This was also true in the Noble Alloy group. In the High Noble Alloy group, Chamfer Finish Lines showed the largest gap when compared to the other two finish lines.

Table 6. (Mean of the internal and marginal fit of the materials with finish lines.)

	Material	Finish Line	Mean	Std. Deviation	N
Internal Fit	Base Alloy	Deep Chamfer	123.2940	20.53320	10
		Chamfer	122.4940	9.83070	10
		Shoulder	113.9340	8.68592	10
		Total	119.9073	14.24140	30
	Noble Alloy	Deep Chamfer	87.7670	13.29221	10
		Chamfer	93.5390	16.17431	10
		Shoulder	87.2050	13.26173	10
		Total	89.5037	14.10957	30
	High Noble Alloy	Deep Chamfer	158.6840	19.20336	10
		Chamfer	149.4950	8.78947	10
		Shoulder	151.3100	23.63601	10
		Total	153.1630	18.11444	30
	Total	Deep Chamfer	123.2483	34.16477	30
		Chamfer	121.8427	25.98493	30
		Shoulder	117.4830	31.08720	30
		Total	120.8580	30.35358	90
Marginal Fit	Base Alloy	Deep Chamfer	19.8000	12.10314	10
		Chamfer	34.8920	10.55894	10
		Shoulder	33.6250	10.25573	10
		Total	29.4390	12.69007	30
	Noble Alloy	Deep Chamfer	34.4990	12.41087	10
		Chamfer	58.5170	10.97184	10
		Shoulder	43.7410	11.64719	10
		Total	45.5857	15.11562	30
	High Noble Alloy	Deep Chamfer	32.3080	12.83968	10
		Chamfer	51.5430	15.56549	10
		Shoulder	46.0500	9.32671	10
		Total	43.3003	14.86781	30
	Total	Deep Chamfer	28.8690	13.70384	30
		Chamfer	48.3173	15.77229	30
		Shoulder	41.1387	11.48298	30
		Total	39.4417	15.82464	90

Table 7. (Multiple comparisons between the materials by post hoc tests.)

Variable	(I) Material	(J) Material	Mean Difference (I-J)	Std. Error	Sig.
Internal Fit	Base Alloy	Noble Alloy	30.4037*	4.04836	.000
		High Noble Alloy	-33.2557*	4.04836	.000
	Noble Alloy	Base Alloy	-30.4037*	4.04836	.000
		High Noble Alloy	-63.6593*	4.04836	.000
	High Noble Alloy	Base Alloy	33.2557*	4.04836	.000
		Noble Alloy	63.6593*	4.04836	.000
Marginal Fit	Base Alloy	Noble Alloy	-16.1467*	3.06395	.000
		High Noble Alloy	-13.8613*	3.06395	.000
	Noble Alloy	Base Alloy	16.1467*	3.06395	.000
		High Noble Alloy	2.2853	3.06395	.737
	High Noble Alloy	Base Alloy	13.8613*	3.06395	.000
		Noble Alloy	-2.2853	3.06395	.737

**P Value ≤ 0.05*

Table 8. (Internal fit corresponding to the materials in Tukey's HSD.)

Material	N	Subset 1	Subset 2	Subset 3
Noble Alloy	30	89.5037		
Base Alloy	30		119.9073	
High Noble Alloy	30			153.1630
Sig.		1.000	1.000	1.000

Table 9. (Marginal fit corresponding to the material in Tukey's HSD.)

Material	N	Subset 1	Subset 2
Base Alloy	30	29.4390	
High Noble Alloy	30		43.3003
Noble Alloy	30		45.5857
Sig.		1.000	.737

Multiple comparisons between the materials using Post Hoc Tests and Tukey Test revealed significant differences between the 3 materials as shown in Table 7.

Regarding the Internal Fit, the highest mean difference was found in the High Noble Alloy followed by the Base alloy. The least mean difference was noted in the Noble Alloy group. All the groups demonstrated significant differences as shown in Table 8.

Considering the Marginal Fit, there were significant differences between the groups except between the Noble Alloy and the High Noble Alloy groups as shown in Table 9.

Table 10. (Mean of the internal fit corresponding to the finish line in Tukey's HSD.)

Finish Line	N	Subset 1
Shoulder	30	117.4830
Chamfer	30	121.8427
Deep Chamfer	30	123.2483
Sig.		.333

Results shown in (Table 10) show no significant differences in all the groups regarding Internal Fit of the different finish lines.

Table 11. (Mean of the marginal fit corresponding to finish lines in Tukey's HSD.)

Finish Line	N	Subset 1	Subset 2
Deep Chamfer	30	28.8690	
Shoulder	30		41.1387
Chamfer	30		48.3173
Sig.		1.000	.056

Results shown in (Table 11) showed a significant difference between the Deep Chamfer Finish Line group as compared to the Chamfer Finish Line and Shoulder Finish Line groups. There was no significant difference between the Chamfer Finish Line Group when compared to the Shoulder Finish Line group.

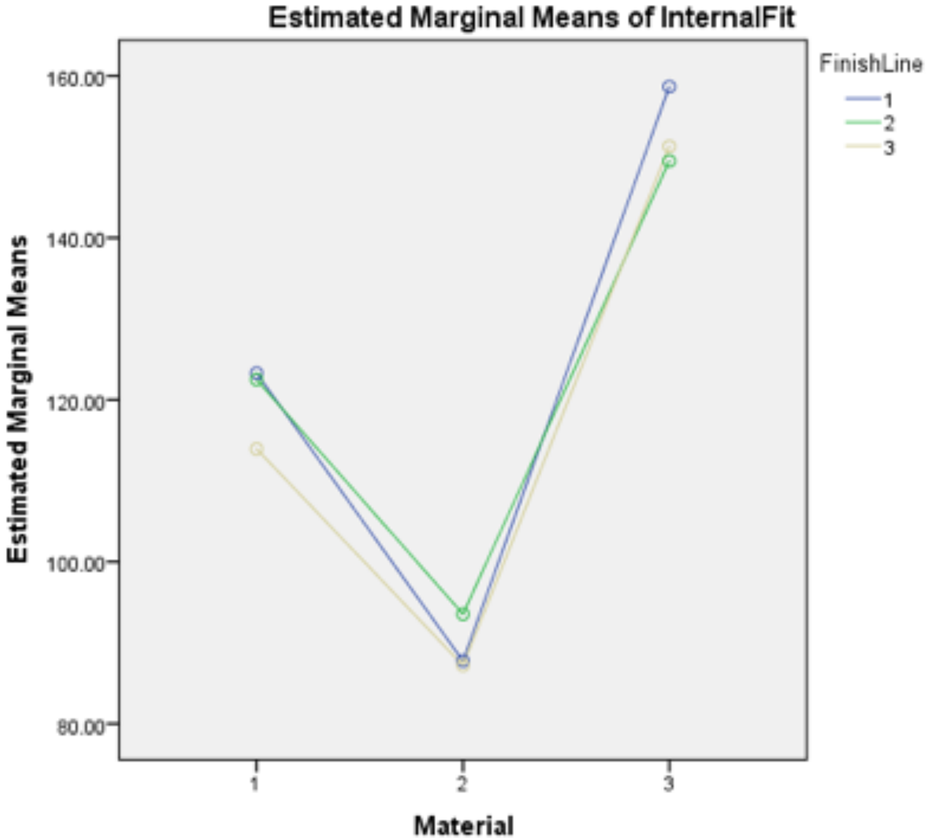

Figure 17. (Estimated Marginal Means of Internal Fit)

** Materials; 1 (Base Alloy), 2 (Noble Alloy), 3 (High Noble Alloy) * Finish Lines; 1 (Deep Chamfer), 2 (Chamfer), 3 (Shoulder).*

As shown in Figure 17, Internal Fit showed the highest gap in the Deep Chamfer Finish Line with the High Noble Alloy group. The smallest gaps were noted in the Shoulder Finish Line group with the Noble Alloy group.

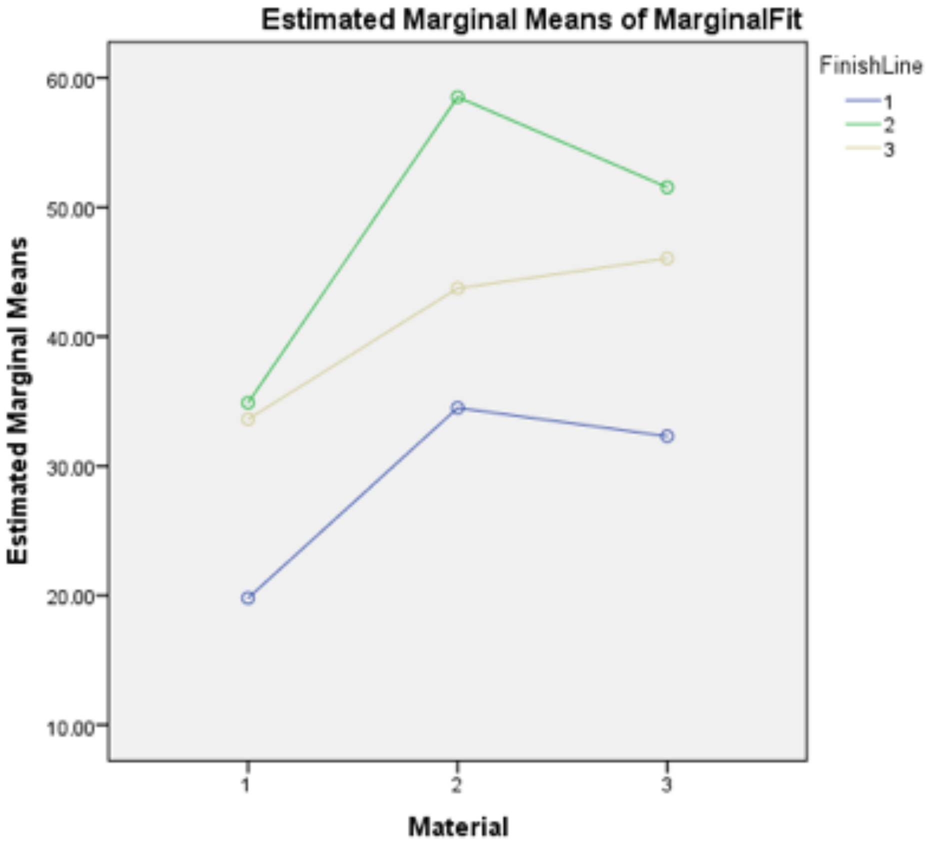

Figure 18. (Estimated Marginal Means of Marginal Fit)

** Materials; 1 (Base Alloy), 2 (Noble Alloy), 3 (High Noble Alloy) * Finish Lines; 1 (Deep Chamfer), 2 (Chamfer), 3 (Shoulder).*

As shown in Figure 18, Marginal Fit showed the highest gap between the Chamfer Finish Line with the Noble Alloy Group. The smallest gaps were noted between the Deep Chamfer Finish Line and Base Alloy Group.

Copings fabricated utilizing Selective Laser Melting (SLM) technology from three different types of alloys yielded a comparable fit. They demonstrated a mean marginal gap in the range of (29-45) µm and an Internal gap in the range of (89-153) µm irrespective to the Finish Line used.

The result of statistical analyses indicated that both alloy type and finish line had a significance influence on overall fit of the copings. For the internal fit, the alloy type had a significant effect ($p<0.001$), but the finish line had no statistically significant influence (p=0.337). For the marginal fit, both the alloy type and the finish line had a statistically significant effect, ($p<0.001$). There was no statistically significant interaction between variables.

For all the finish lines used, the lowest marginal gaps were obtained in the Base Alloy group (29) μm. No statistical significant differences existed among the High Noble Alloy group (43) μm and the Noble Alloy groups (45) μm.

Regarding the internal fit of the three different alloy groups irrespective to the Finish Line used, there were significant differences among the groups. The lowest internal gap was in the Noble Alloy group (89) μm followed by the Base Alloy group (120) μm and High Noble Alloy group (153) μm.

Considering the finish lines without considering the Alloys used, the mean values for the internal fit measurement were (123) μm, (122) μm, and (117) μm for the Deep Chamfer, Chamfer, and Shoulder finish lines respectively. There were no significant differences between the 3 mean values.

The marginal gap was (48) μm for the Chamfer Finish Line group, (41) μm for the Shoulder Finish Line group and (29) μm for the Deep Chamfer Finish Line group. According to the results of this study, the best finish line design was the Deep Chamfer Finish line irrespective to the alloy used. There were no significant differences between Chamfer Finish Line and Deep Chamfer Finish line groups.

The Internal fit of the Noble Alloys group with the Deep Chamfer Finish lines and Shoulder Finish Lines showed the smallest internal gap (88 ± 13) μm and (87 ± 13) μm, respectively. Whereas copings made with High Noble Alloys and Deep Chamfer Finish lines showed the largest internal gap (159 ± 19) μm.

The marginal fit of the copings in the Base Alloys group with the Deep Chamfer Finish lines had the best marginal fit (20 ± 12) μm. Whereas copings made with the Noble alloy and with Chamfer Finish Lines showed the least acceptable marginal fit (59 ± 11) μm.

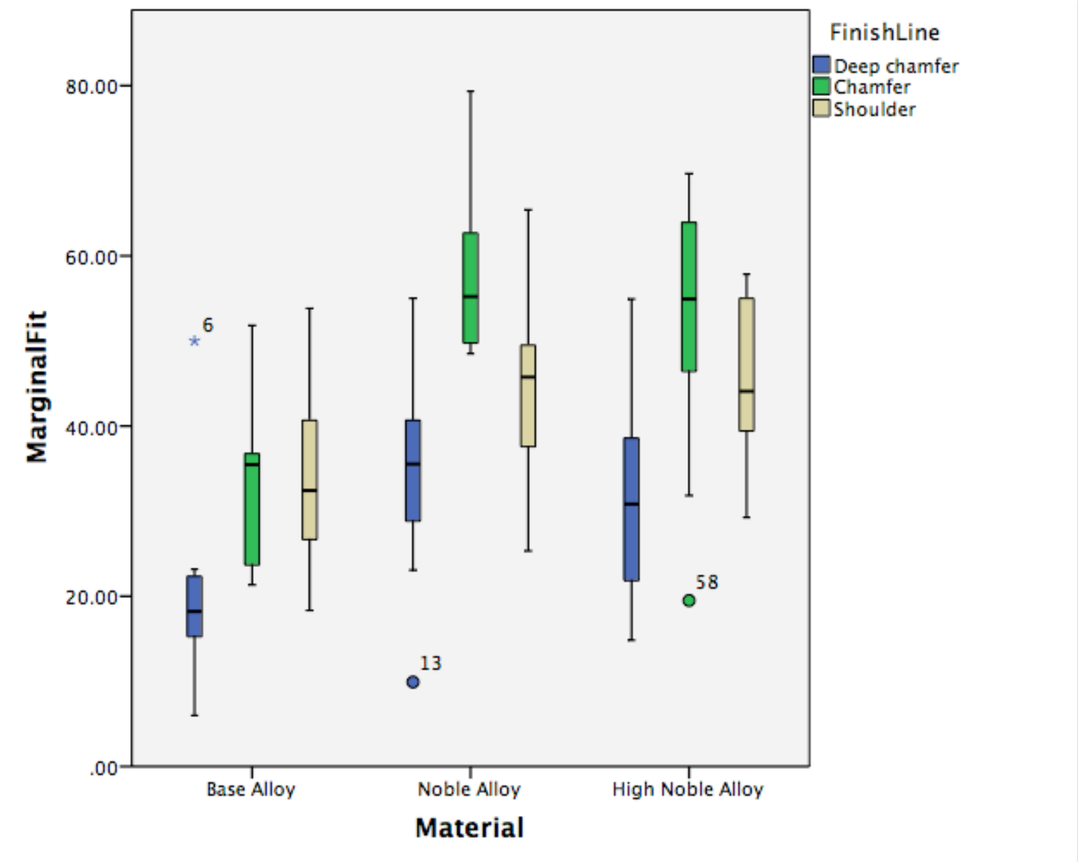

Figure 19. (Boxplot (mean of marginal fit).)

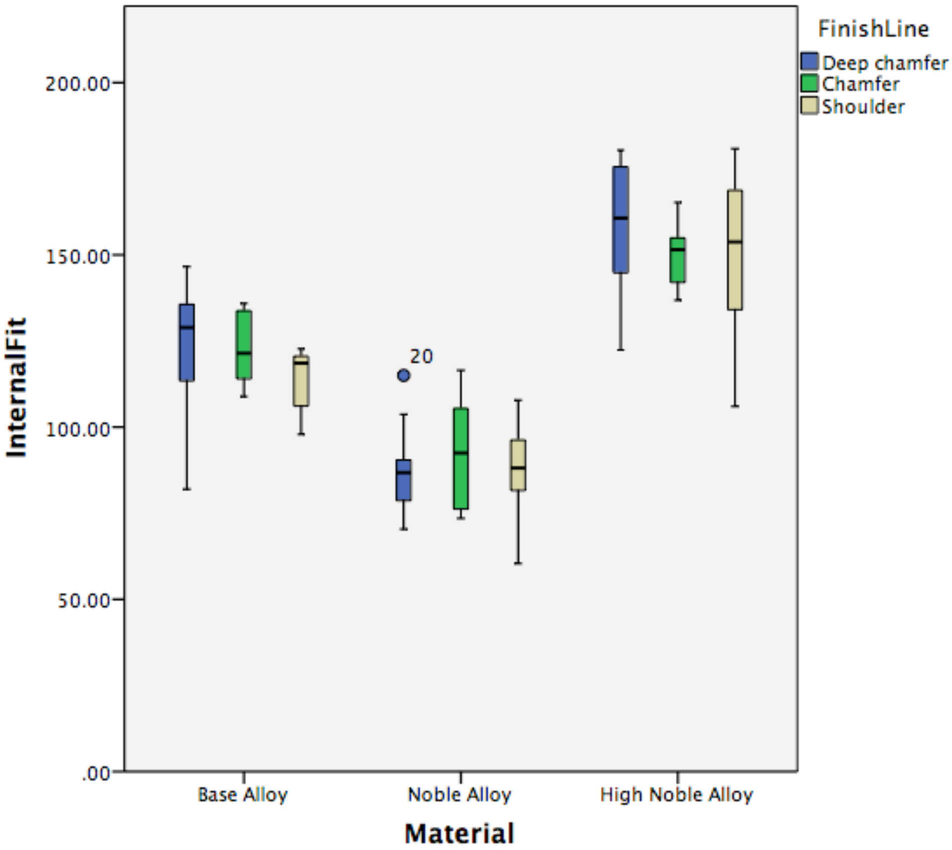

Figure 20. (Boxplot (Mean of Internal Fit).)

<u>Microscopic Images</u>

Overall, 2700 measurements (30 measurements × 90 specimens) were obtained for the study. The microscope was linked to a digital acquisition device and computer software (Spot Software 5.2, Spot Imaging Solutions).

Below are representative microscopic images showing different alloys with different finish lines and measured at different locations.

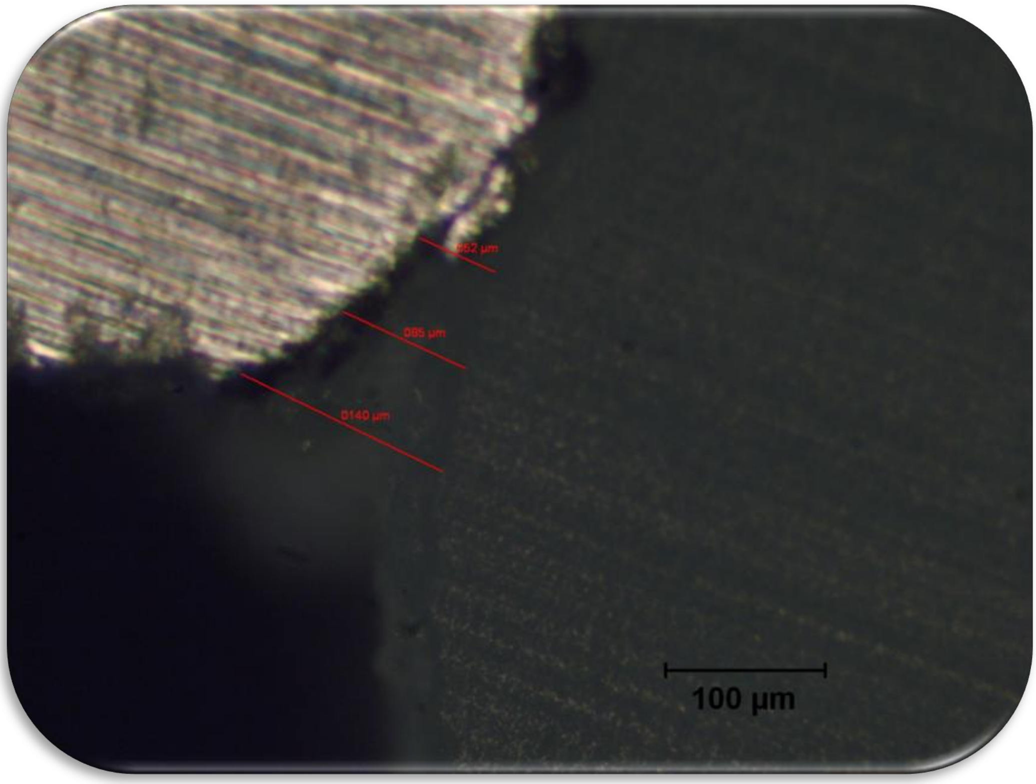

Figure 21. (Noble alloy; facial margin, chamfer finish line.)

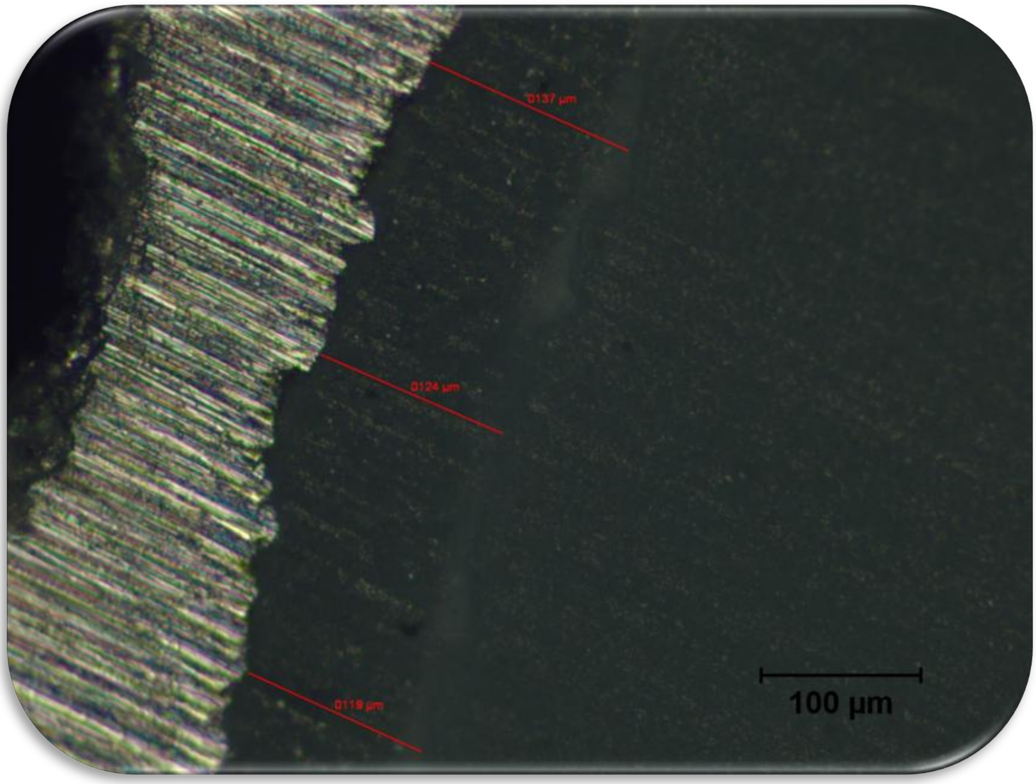

Figure 22. (Base alloy; facial midaxial, deep chamfer finish line.)

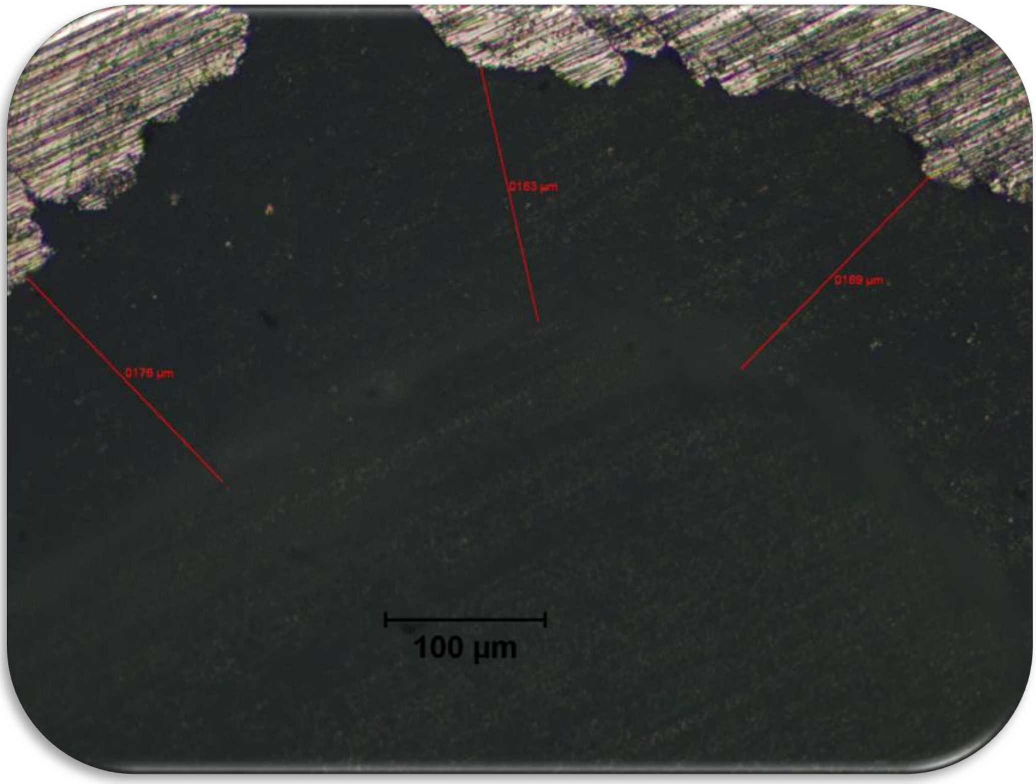

Figure 23. (Base alloy; incisal, chamfer finish line.)

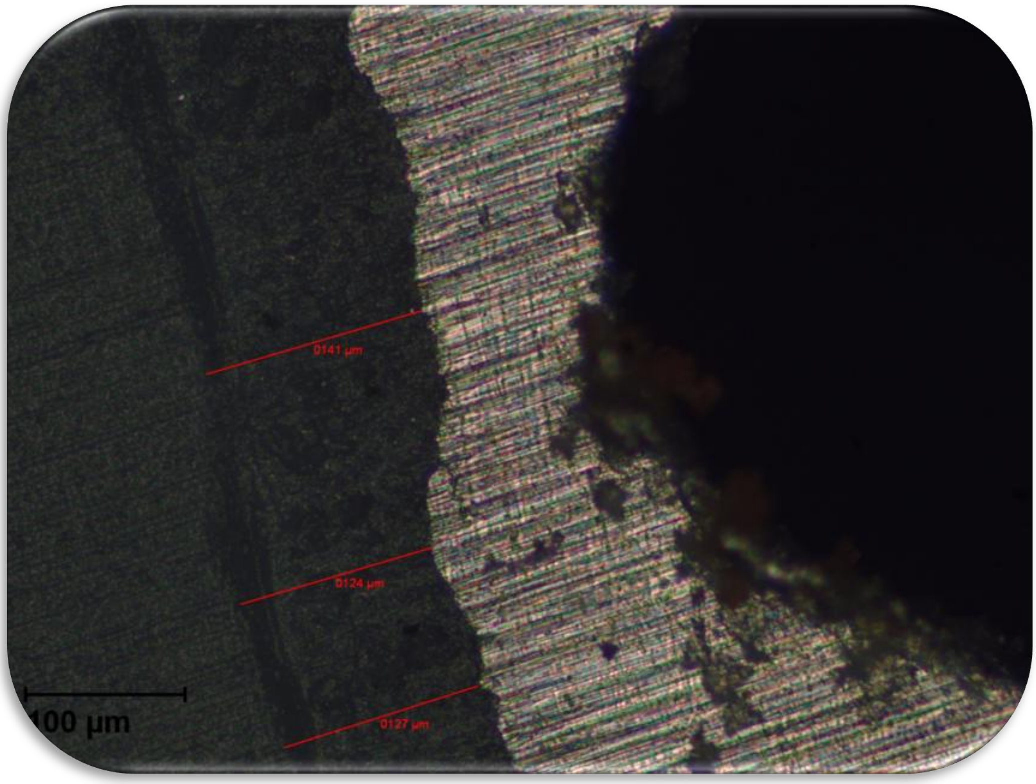

Figure 24. (Base alloy; lingual midaxial, shoulder finish line.)

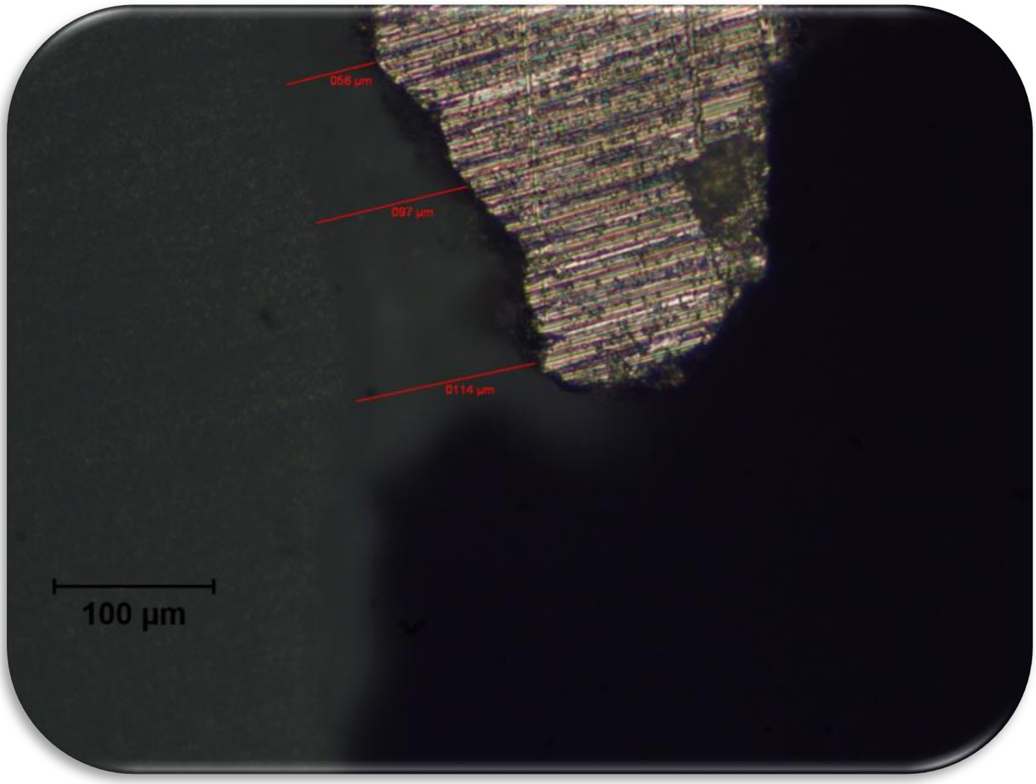

Figure 25. (High noble alloy; lingual margin, chamfer finish line.)

CHAPTER VI

DISCUSSION

The primary purpose of this study was to evaluate the effect of 3 different finish line designs with 3 different alloys on the marginal and internal fits of SLM-fabricated copings. The first null hypothesis was rejected as statistically significant results were found among the 3 finish line groups. Therefore, the type of finish line design had a direct effect on marginal gaps noted between the copings and the dies. However, no statistical difference was found on the internal fit relative to the different type of finish lines used. The second null hypothesis was also rejected as there were statistical differences between the type of alloys used and the marginal and internal fit of the SLM-fabricated copings.

Terminology varies when it comes to defining the word "fit". The same term has been used to describe multiple different measurements. There are no clear general guidelines for performing gap measurements of dental restorations. Holmes et al. established a critical approach to this problem.(7) They established multiple gap definitions according to the contour differences between crown and tooth margins. Nevertheless, in clinical practice, it has been extremely difficult to describe a gap using only a definition due to morphologic diversities, rounded margins or defects.(107) This is the main reason why many investigators report widely different results when it comes to measuring gaps of crown/tooth marginal gaps.

In this study, according to Myung-Joo Kim et al., marginal gaps were defined as "two dimensional vertical marginal discrepancy measured from the coping to the margin of the preparations". Internal gaps, per according Myung-Joo

Kim et al., were defined as the "vertical measurement from the internal surface of the copings to the axial walls of the preparations".(108)

Multiple techniques have been advocated for measuring the marginal and internal fit of crowns. Direct viewing, cross-sectional, impression technique, explorer and visual examination have been used most often.(109) In this study, the cross-sectional technique was used after the cementation of the copings onto the dies as it had multiple advantages over previously cited techniques.

Sorensen described the cross-sectional method to measure marginal accuracy. Sorensen's technique permitted a comparison of different margin designs and the evaluation of the fit of restorations. Although this technique is time consuming and required many steps, it also resulted in significant waste of laboratory specimens (crowns). It did provide more information and greater precision of measurement than other modalities. The cross-sectional evaluation of the margins permitted more precise measurement of predetermined points which was not possible with the direct viewing technique.(109)

In this study, the material used to fabricate the dies was a 3D printed DPR 10 Resin (Carbon3D, USA). Several investigation have used metal, acrylic resin or natural teeth to measure the marginal fit between crowns and preparations.(110–114) The advantages of DPR 10 printed dies that were used in this study are the standardization of all the copings, and the ability to print as many dies as necessary for the study without having a large discrepancy between the specimens. Moreover, there is a lack of wear during the fitting process and improved fitting accuracy.

Die spacing methods have specific differences for each system and can influence the fit of the restorations.(104) Weaver et al. found that the amount of die spacer used had a specific factor for fit.(115) Therefore, in the present study, die spacer was not manually applied on the surfaces of the dies, rather it was specified during the design process of the copings using 3Shape CAD design software. The advantage of using software to determine the amount of die spacer used eliminated the differences that can occur depending on the practitioner applying the die spacer.

Well-fitted metal copings during the try-in phase might not fit accurately after porcelain application.(116) Anusavice et al. believed that the majority of the changes in the alloy occurred during the oxidation cycle.(117) Campbell et al. and Gemalmaz et al. reported in their respective studies, that marginal gaps increased significantly following ceramic application.(118,119) On the contrary, multiple studies found no significant differences on the marginal gaps before and after ceramic application on restorations.(109,116,120,121)

In an effort to make the measurements as accurate as possible, and to focus on SLM-fabricated copings' marginal and internal fits, this study measured the cemented copings without porcelain veneering so as to not complicate the results with other variables and factors. Sulaiman et al. and Beschnidt SM et al. used a similar technique to determine the coping fit without application of porcelain to the copings.(122,123)

Even though statistically significant differences of the marginal fit occurred between the different types of finish lines used with different types of alloys used in the present study which ranged from (20 to 59) µm, the results were found to be with in clinically acceptable levels. McLean and Von Fraunhofer in a clinical study of 100 restorations over a 5-year period, hypothesized that (120) µm represented the maximum clinically acceptable misfit.(6)

Bindl and Mormann reported acceptable internal gap widths of (81 to 136) µm for different all-ceramic CAD/CAM crown copings. These findings reported gap measurements that were greater, for two of the alloys (noble and base metal) but less than the gaps noted (153) μm with the high noble alloy group recorded in the present study.(1)

Katrin et al. studied the marginal and internal fit of precious and base alloys fabricated with laser melting technology. They found no significant differences in marginal discrepancies and internal fits between the two types of alloys.(20) The results of Katrin et al. contradicted the results of the present study, where it was found that the type of alloy did have a significant difference on marginal and internal fit of the SLM-fabricated copings.

As the concept of minimally invasive dentistry is spreading, more clinicians are willing to implement that principle in their practice.(124,125) However, as the minimal preparation design is highly preferred, there might be some constraints on the tooth design by the material used and its method of fabrication.(100) In this present study, SLM technology clearly showed less capability for capturing the chamfer finish line preparations when compared to heavy chamfer or a shoulder finish line preparations with all the different types of alloys used in this study.

In this study, it was found that the marginal fit of the copings fabricated with Base Alloy (Co-Cr) and deep chamfer finish lines had the best marginal fit of (20 ± 12) μm. This fact leads the authors to believe that Co-Cr alloy crowns made by SLM technology could result in widespread clinical use, even though its present use is limited. Research on surfaces of SLM-fabricated Co-Cr alloys crowns have demonstrated that they have rougher surfaces than those made by conventional casting procedures with the same composition. This has an advantage over conventional castings because it positively affects the metal ceramic bond. It is of interest that the composition of the Co-Cr alloy used in this study for SLM did not contain tungsten and had a lower molybdenum content when compared to the composition of Co-Cr alloys for casting. Ucar et al. presumed “laser sintering of the former Co-Cr alloy is facilitated by the absence or diminished percentage of such refractory metals, which have much higher melting temperatures than cobalt and chromium”.(126)

In the present study, marginal fit was influenced by the type of finish line; deep chamfer finish lines were better when compared to the marginal gaps associated with chamfer and shoulder finish lines. It was not in agreement with the results of Zen et al’s study, as they found that marginal fit was not influenced by the type of finish line in the preparations.(127)

Limitations of the Study

The limitations of this study include that the assessment of marginal and internal fit were not performed intraorally and that the errors in fabrication and handling of dies were assumed to be minimal. Further studies are required for clinical application and assessment of the present data. Future research should include biocompatibility of restorations prepared by selective laser melting (SLM) technology.

Another limitation of the study was that only copings were fabricated using SLM; therefore, the influence of porcelain firing on the marginal and internal fit of the crowns was not measured.

The copings fabricated in this study were not subjected to mechanical and thermal cycling. It is well known that thermo-mechanical cycling may be one of the important factors that affect the long –term success of the restorations and may have an impact on accuracy of marginal and internal fit of SLM-fabricated copings.

CHAPTER VII

CONCLUSIONS

Within the limitations of the study, the following conclusions were drawn:

1. The automatic fabrication process resulted in accurate marginal and internal fits of the SLM-fabricated copings and minimized errors due to casting shrinkage and human errors.
2. Coping fabricated with this SLM technology fit within pre-established, clinically acceptable ranges.
3. Finish line configurations and alloys used in this study influenced the marginal fit of the SLM-fabricated copings.
4. Finish line types did not significantly influence the internal fit between the copings and the dies, whereas alloy type did influence the fit between copings and dies.
5. SLM-fabricated copings made with the Base Alloy (Co-Cr) on teeth prepared with deep chamfer finish lines demonstrated the best marginal fits when compared to the other groups.

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