Retentive force and fitness accuracy of cobalt-chrome alloy clasps for removable partial denture fabricated with SLM technique

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Abstract

Purpose: Evaluating the fitness accuracy and retentive force of cobalt-chrome (Co-Cr) alloy clasps fabricated using the selective laser melting (SLM) technique.

Methods: Premolar and molar abutment models with a 0.5-mm undercut depth, 1.5-mm-thick occlusal rest seats, and guiding planes were designed and fabricated using a milling machine. On these models, Akers clasps with 0.25- and 0.5-mm undercut depths were designed and fabricated with SLM and a traditional lost wax casting method. Based on the manufacturing methods, abutment types, and undercut depths, the clasps were divided into eight groups (10 per group). The fitness accuracy of the clasps was evaluated by measuring the gap distance between the clasps and abutments using a silicone film method. The initial retentive force and changes in retention up to 7,200 insertion/removal cycles of the clasps were also measured. The data were analyzed using multiple linear regression, paired t-tests, and one-way ANOVA (α=0.05).

Results: For both the SLM and cast clasps, the fitness accuracy of the rest was greater than that of the clasp tip and shoulder. No significant difference was found in the fitness accuracy between the SLM and cast clasps, regardless of the abutment type and undercut depth before or after insertion/removal cycles (p>0.05). There was also no significant difference in the initial retentive force between the SLM and cast clasps (p>0.05). After 7,200 insertion/removal cycles, the SLM clasp exhibited a greater residual retentive force (p<0.05).

Conclusion: The SLM technique for manufacturing the clasps of removable partial dentures has promising clinical applications.

Keywords: SLM, Clasp, Retentive force, Fitness accuracy, Cobalt-chrome alloy

1. Introduction

Removable partial dentures (RPDs) are patient-specific prostheses with complex geometries that are conventionally fabricated using a lost-wax casting procedure. The procedure involves several steps that require the skilled hands of a technician; thus, the entire process is time-consuming. Recently, selective laser melting (SLM) has been introduced to fabricate RPDs and has attracted much attention[1–3]. SLM is an additive manufacturing (AM) technique based on layer-wise material addition that allows the generation of complex 3D parts by selectively melting successive layers of metal powder on top of each other. Previous studies have shown higher yield strength, ultimate tensile strength, fatigue resistance, and corrosion resistance of a cobalt-chromium framework processed by SLM over a lost-wax casting procedure[4–8]. Therefore, the RPD represents an application in which SLM may be particularly advantageous.

Some in vivo and in vitro studies have evaluated the fitness accuracy and retentive force of RPDs. Schweiger et al. studied the retentive force of the embrasure clasps after 65,000 cycles of simulated ageing and revealed that the residual retentive force of SLM clasps were considerably higher than those of the cast clasps[14]. Nakata et al. and Torii et al. tested the fitness and retentive force of cobalt-chrome (Co-Cr) alloy clasps using one-process molding by repeated laser sintering and high-speed milling on a simple simulated molar model in distilled water, showing a similar fitness accuracy of the SLM clasp and the cast clasp in the arm and tip regions. They also found that the long-term retentive force of the SLM clasp was superior to that of the cast clasp[15,16]. Torii et al. improved the design of the Akers clasp by providing 50 μm of digital relief on the occlusal surface of the tooth die, which resulted in better fitness accuracy and...
improved retentive force for digital clasps compared to conventional cast clasps[16].

Although the hybrid machine with a one-process molding of laser sintering and milling is convenient and economical, machines using SLM alone are still relatively common. Therefore, in the present study, we investigated the fitness accuracy and retentive force of a Co-Cr clasp made using the SLM technique on models with a size equal to that of the natural teeth, including premolars and molars, as well as in artificial saliva, to simulate the oral environment.

2. Materials and methods

2.1. Design and fabrication of abutment models

Standard first premolar and first molar digital data of the right lower jaw were acquired from the database of the dental design software (3Shape’s 3rd Generation Dental System, 3Shape, Denmark) and were modified to form a type I guide line with 0.5-mm undercut depth. A rest seat was prepared on the occlusal surface, with a depth of 1.5 mm, a width of 1/3 of the buccolingual dimension of the molar and 1/2 of the buccolingual dimension of the premolar[17]. A guide plane was prepared on the proximal and lingual surfaces[18]. The teeth were then added to a cubic base (10 mm×10 mm×10 mm) using a CAD system (Geomagic Studio 12.0, Geomagic Software, USA). The lateral face of the cubic base is parallel to the guide plane. With these two designed digital data sets, five titanium abutment models of the first premolars and first molars were each milled using a milling machine (JK20-5AXIS, Runyes, China) without polishing (Fig. 1).

2.2. Design and fabrication of clasps

According to the manufacturing method, abutment tooth type, and undercut depth, the clasps were divided into eight groups (expressed as method-tooth-undercut): cast-46-0.25, cast-46-0.5, cast-44-0.25, cast-44-0.5, SLM-46-0.25, SLM-46-0.5, SLM-44-0.25, and SLM-44-0.5. Cast Co-Cr clasps were used as controls. The clasps were designed using dental design software (3Shape’s 3rd Generation Dental System, 3Shape, Denmark). The Akers clasp was designed such that the buccal tip of 1/3 of the retentive arms was engaged at the 0.25-mm and 0.5-mm undercuts. Then, the occlusal rests, clasp bodies, and guiding plates were successively added to the clasp arms. The thickness and width of the clasp tip were 0.8 and 1.6 mm, respectively[19]. The length of the clasp arm was 15 mm for tooth 46 and 11 mm for tooth 44. The cross section of the clasp was a semicircle. A holding bar that was 3 mm in diameter, 25 mm in length, and parallel to the insertion path was added to the occlusal rest using Geomagic software. STL clasp (Fig. 2A) data were sent to the SLM machine (M2 cusing Multilaser, Concept Laser, Germany), and Co-Cr alloy powder (Wirobond C+, Bego Dental, Germany) was used to fabricate the SLM clasp. The SLM parameters were as follows: laser power: 160 W, scanning speed: 800 mm/s, layer thickness: 0.03 mm, and the building direction was perpendicular to the holding bar. The SLM clasps were then heated from 650 °C to 800 °C within 12 min. The temperature was held for 15 min, followed by cooling to 550 °C within 15 min for stress relief of clasps, according to the manufacturer’s instructions. To ensure the consistency of the cast clasp with the SLM clasp, STL clasp data were also used to fabricate the resin pattern using a 3D printer (ProJet MJP 3600, 3D Systems, America) for the cast clasp. The tip of the cast clasp arm was a closed circuit (Fig. 2B) to avoid deformation during casting. The refractory mold was duplicated from the titanium abutment model using the silicone rubber method, and the resin pattern was then used to fabricate the cast clasp by a traditional lost-wax casting method using Co-Cr alloy metal blocks (Wironit, Bego Dental, Germany). The compositions of the two alloys provided by the manufacturers are listed in Table 1. To ensure the uniformity of the clasps, neither the cast nor SLM clasp specimens were polished. Each of them was only airborne-particle abraded with 50-μm aluminum oxide under 0.5 MPa air pressure for 30 s[15]. The support, nodules, and burs were carefully removed. Finally, 10 clasps were completed in each of the eight groups.
2.3. Microstructural observation

SLM and cast alloy block samples (10 mm×10 mm×2 mm) were prepared in the same manner as described above. After etching in hydrochloric acid and hydrogen peroxide (4:1, v/v) for 30 s[4], they were observed by scanning electron microscopy (SEM, S-3400N, Hitachi, Japan) to characterize the grain size and morphology.

2.4. Non-destructive inspection

Before the measurements, a non-destructive inspection of all clasps was performed by taking radiographs using an X-ray generator (Heliodent Plus D3507, Sirona, Germany) with an exposure time of 0.64 s and a distance of 10 cm between the focus and the film.

2.5. Measurements of fitness accuracy

The evaluation of the fitness accuracy of the clasps was performed by measuring the gap distance between the clasps and abutment dies using the silicone film method in accordance with the procedures established in previous studies[20–22]. A high-viscosity silicone impression material (Express XT light body, 3M, America) was mixed and applied on the surface of the die, and the clasp was then seated on it. The clasps were held under a constant load of 9.80 N for five minutes and then carefully removed from the die. To avoid the deformation and breaking of the silicone film, the orange silicone material (Express STD firmer set, 3M, America) was mixed with a catalyst and injected onto the pink silicone material of each clasp. Five minutes later, the two combined silicone materials were removed from the clasp. The silicone block materials from each clasp were sectioned with a razor blade buccal-lingually 0.5 mm from the end of the clasp tips, clasp shoulders, and rest regions. The pink silicone layer at the sections was measured to determine the clasp fitness accuracy using a stereo microscope (SteREO Discovery.V12, Carl Zeiss, Germany) at a magnification of 100×. These measurements were also performed after 7,200 insertion/removal cycles.

2.6. Measurement of retentive forces

A tensile test apparatus (HY-0230, Hengyi, China) was used to measure the retentive force. One of the premolar abutments and one of the molar abutments was selected as the abutment PA and MA, which were only used for the measurement of the retentive force. First, PA and MA were engaged with clasps and immersed in Fusayama-Meyer artificial saliva, and the holding bar was connected to the apparatus (Fig. 4). The Fusayama-Meyer artificial saliva solution was prepared from deionized water and the following analytic grade reagents: 0.4 g/L NaCl, 0.4 g/L KCl, 0.795 g/L CaCl2·2H2O, 0.690 g/L NaH2PO4·H2O, 0.005 g/L Na2S·9H2O, and 1.0 g/L urea[4]. The initial retentive force of each clasp was measured as the maximum tensile force obtained when the clasp was separated from the die at a speed of 50 mm/min. The measurement was repeated five times, and the average values were calculated as the initial retentive force.

The apparatus shown in Fig. 5 was used for the insertion/removal cycles, which was designed at the Shanghai Jiao Tong University School of Medicine. Each group of clasps had its own abutment model for each cycle. After each clasp was engaged in the abutment die and immersed in Fusayama-Meyer artificial saliva, it was inserted into the terminal position of the die and subsequently separated from the die to simulate the insertion and dislodgement of dentures. These actions constituted one cycle. Repeated tensile and compressive motions (crosshead speed: 45 cycles/min) were performed for up to 7,200 cycles to simulate the clinical use of the clasp in the mouth for 5 years[18]. After each set of 360 cycles (i.e., three months) and after the final 7,200 cycles, the clasps were returned to the abutment PA or MA, and the retentive force of each clasp was measured according to the aforementioned method. The fracture of the clasp was regarded as a fatigue failure, and the cycle number was recorded.

2.7. Statistical analysis

The fitness accuracy and retentive force data between different groups were analyzed using multiple linear regression with the SPSS statistical package (SPSS Statistics 20, IBM, USA). The fitness accuracy and retentive force of the clasps before and after the cycles were compared using a paired t-test. The fitness accuracy data of the clasp tip, clasp shoulders, and rest regions were compared using one-way ANOVA and Tukey’s multiple comparison test. The statistical significance was set at α=0.05.

3. Results

3.1. Microstructure

The microstructures of the cast and SLM sample surfaces are shown in Fig. 6. The cast sample consisted of coarse dendrites with visible precipitates in the inter-dendritic regions, and the diameter of the precipitates was approximately 5-10 μm. In contrast, cellular dendritic structures, with a diameter of approximately 0.5 μm at a magnification of 10,000 ×, were observed in the SLM sample.

3.2. Fitness accuracy of clasps

The initial fitness accuracy is shown in Fig. 7. The fitness accuracies of each part of the clasp ranged from 28.6 μm to 155.9 μm in the SLM group and from 23.3 μm to 150.5 μm in the cast group. Regardless of the manufacturing method, type of abutment tooth, and undercut depth, there were no significant differences in the gap distance at the region of the clasp tip and shoulder (p>0.05). Conversely, in the region of the occlusal rest, for both the cast clasp and SLM clasp, the gap distance of the first molar was greater than that of the first premolar (p<0.05). In addition, the gap distance in the region of the
occlusal rest was greater than that in the region of the clasp tip and shoulder (p<0.05).

The final fitness accuracy after 7,200 cycles is shown in Fig. 8. For each group of clasps, the gap distance at the region of the clasp tip increased significantly compared to the initial value (p<0.05), but there were no differences in the region of the occlusal rest and the clasp shoulder (p>0.05). For the same tooth position and undercut, the gap distance at the region of the clasp tip was smaller for the SLM than for the cast, but the difference was not significant (p>0.05).

3.3. Retentive forces of clasps

The retentive forces before and after the cycles are listed in Table 2. At the same tooth position and undercut, there was no difference in the initial retentive force between the SLM clasp and cast clasp (p>0.05). For the same tooth position and manufacturing method, the initial retentive force of the 0.5-mm undercut was greater than that of the 0.25-mm undercut (p<0.05).

After 7,200 insertion/removal cycles, no fractures were observed in any of the 80 clasps. For each group of clasps, the retentive force significantly decreased (p<0.05). At the same tooth position and undercut, the SLM clasp showed a lower decrease in retentive force compared with the cast clasp (p<0.05). For the same undercut and manufacturing method, group 46 showed a lower decrease in retentive force compared with that of group 44 (p<0.05).

The changes in the retentive force during the insertion/removal cycles are shown in Figs. 9-10. In each group of clasps, the retentive force decreased along with the cycle number, and the most significant decrease occurred in the first and second 360 cycles, which simulated the clinical use of clasps for approximately 3-6 months. After the second 360 cycles, the decrease became relatively stable. The tendency towards change in the retentive force was almost the same regardless of the clasp type, tooth position, and undercut depth.

4. Discussion

SLM technology has recently become popular in the field of oral medicine and has many advantages, such as rapid prototyping and offering savings in materials and time[15]. In the present study, we investigated the fitness accuracy and retentive force of the Co-Cr clasp made by the SLM technique, which is very important for achieving a desirable function[15]. The results showed that the fitness accuracy of the SLM clasp ranged from 28.6 µm to 155.9 µm, nearly the same as that of the cast clasp. Previous studies have reported that the clinically acceptable values of fitness accuracy for frameworks fabricated by conventional casting were 69-387 µm[23,24]. Another systematic review showed that the average fitness accuracy of SLM-produced frameworks is within the range of 50-380 µm[8], which is consistent
with our results. Therefore, the fitness accuracy of the SLM clasp was clinically acceptable.

Nakata T et al. tested the fitness accuracy of the Co-Cr clasp using one-process molding by repeated laser sintering and high-speed milling (HM) on a simple simulated molar model and showed that the gap distance at rest, the clasp arm, and the clasp tip were 167.4, 63.4, and 61.0 μm, respectively[15]. Our study also showed that, for SLM clasps, the average gap distance at the clasp arm was 57.8 μm and the clasp tip gap distance was 50.9 μm, which were quite similar to those of the HM clasps. However, in the rest region, the average fitness accuracy of our SLM clasp was 91.1 μm, which was better than that of the HM clasp. This is because in their study, the shifting corner of the rest and the clasp body was slightly acute. Thus, the corners can-
not be accurately milled[15]. To solve this problem, Torii et al. used the same machine and improved the design of the Akers clasp on a simplified and flat model with 50 μm of digital relief on the occlusal surface of the tooth die, which resulted in better fitness accuracy of the rest region[16]. Here, we chose natural tooth data as an abutment model, which has a more complex tooth surface and inclination. In addition, we selected the SLM technique alone to fabricate clasps. The results showed that the fitness accuracy of the occlusal rest was worse than that of the clasp arm and clasp tip. A similar trend was shown in Huang et al.’s study, which investigated the marginal and occlusal fitness values of the SLM cobalt-chromium alloy crown[25]. It was reported that because of the layer-by-layer production procedure, the external surface tends to have a stepped and coarse morphology representing each fabrication layer along the construction direction[26]. Such stepping adversely affects the surface texture and the overall dimensional accuracy of the workpiece, especially corrugated or sloping surfaces[26]. In addition, Vandenbroucke et al. found that the melting of the enclosed loose powder by surrounding heat during the manufacturing process would result in poor fitness in small and complex structures[27]. As the area of the occlusal rest, the corrugated surfaces and complex structure for group 46 were larger than those for group 44, the adverse effect on fitness accuracy of the occlusal rest for group 46 was greater, resulting in significantly worse fitness at the occlusal rest of 46 compared with 44, although both were within the clinically acceptable range.

The retentive force of the clasp depends on the amount of undercut, arm length and thickness, cross-sectional area, elastic modulus of the clasp materials, and friction coefficient[18,19]. In the present study, the elastic moduli of the SLM and cast clasps were 210 and 211 GPa, respectively, according to the manufacturer’s specifications. The cast and SLM clasps were fabricated with the same digital data such that all the clasps had the same arm length, thickness, and cross-sectional area. The friction coefficient depends on the surface roughness of the material. Takaichi et al. revealed that there is no significant difference in the surface roughness of the clasp arm between the SLM and cast clasps[8]. Given the same level of undercut, the initial retentive force of the clasp is determined by its fitness accuracy. Our results showed that there was no difference in the fitness accuracy between the cast and SLM clasps, which could explain why there was no difference in the initial retentive force between them. Based on previous studies, the required retentive force of one clasp would be 5-10 N to guard against removal while chewing sticky foods and to avoid injuring the abutment tooth[28]. Therefore, when the undercut depth was 0.25 mm, the retentive force of both cast and SLM clasps met the clinical requirements. When the undercut depth was 0.5 mm, the initial retentive force was too high (greater than 10 N). Therefore, for the SLM and cast cobalt-chromium alloy Akers clasps for both premolars and molars, a 0.25-mm undercut depth is reasonable, which is also consistent with the requirements of the clasp set by the traditional method.

After the insertion/removal cycles, obvious reductions in retentive force were observed in each group of clasps. This may be due to two reasons. First, the attrition at the internal surface of the clasp tip and the abutment die, combined with the fatigue deformation of clasps during the cycles, could lead to a greater gap between the clasp and the abutment. Second, attrition also led to a decrease in the friction coefficient, which was caused by a decrease in surface roughness. In this experiment, each group of clasps has its own abutment model for insertion/removal cycles to avoid differences in the degree of wear and the friction coefficient of the abutment in insertion/removal cycles due to different experimental sequences. Additional premolar and molar metal abutments were used to test the retentive force to eliminate the influence of abrasion on the abutment die. The retentive force of the clasp in each group decreased significantly after 360-720 cycles, which is believed to be mainly caused by the rapid decrease in the friction coefficient of the inner surface of the clasps[18].

The present study also showed that SLM samples exhibited higher density surfaces and smaller grain sizes compared with those of cast samples under microscopic observation. Studies have reported that the surface hardness increases with the density of the surface, and a smaller grain size leads to a higher fatigue resistance[4,29,30]. Therefore, the difference in microstructure demonstrated the superiority of the SLM clasp in terms of hardness and fatigue resistance over the cast clasp, which may be the reason why the decrease ratios of fitness accuracy and retentive force of the SLM clasp were smaller than those of the cast clasp after insertion/removal cycles. Alsheghri et al. reported that circumferential clasps with a smaller radius of curvature, such as those in premolars, would experience higher tensile stresses than those in molars and are more susceptible to plastic deformation and fatigue failure[31]. This could be the reason why the decrease ratio of the retentive force for group 44 was greater than that for group 46. The final retentive force of every group was greater than 5 N, except for the cast-44-0.25 group, suggesting that either SLM or the cast clasp can still meet the clinical requirement 5 years later. Additionally, the residual retentive force of the SLM clasp was higher, which was conducive to a high 5-year survival rate for RPDs.

In this study, standard models of natural teeth were adopted and then immersed in Fusayama-Meyer artificial saliva for insertion/removal cycles and retentive force tests to simulate the oral environment as much as possible. However, the real oral environment is more complex and includes different types of bacteria and other organisms. Moreover, the presence of the periodontal membrane in natural teeth also has an effect on retentive force. Therefore, further clinical trials are needed to verify the clinical applicability of the SLM clasp in the future. Furthermore, we found the strange phenomenon that the retentive force decrease ratio of group 0.25 was greater than that of group 0.5 in the present study. A previous study reported that a greater initial retentive force required a greater dishlodging force to remove the clasp from the undercut area, which may cause more fatigue deformation and internal surface attrition of the clasp tip[18]. Theoretically, the retentive force decrease ratio of group 0.5 should be higher. We cannot explain this phenomenon; therefore, further research is necessary in the future.

5. Conclusion

Within the limitations of this study, the following conclusions were reached:

1. There was no significant difference in fitness accuracy between the SLM and cast clasps.

2. The retentive force of the SLM clasp was similar to that of the cast clasp before the cycle testing. Additionally, whether used in 44 or 46 abutments, the undercut depth of 0.25 mm indicated a proper design.

3. After the insertion/removal cycles, the SLM clasp exhibited a greater residual retentive force than the cast clasp and met the clinical requirements for five years.
These results suggest that the SLM technique for manufacturing the clasp of RPDs is promising for clinical applications.

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Conflicts of Interest

There are no conflicts to declare.

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