Cobalt-chromium alloys are used to fabricate various dental prostheses, and have advantages of low cost and excellent mechanical properties compared to other alloys. Recently, selective laser melting, which is an additive manufacturing method, has been used to overcome the disadvantages of the conventional fabrication method. A local rapid heating and cooling process of selective laser melting induces fine microstructures, grain refinement, and reduction of porosities of the alloys. Therefore, it can improve mechanical properties compared to the alloys fabricated by the conventional method. On the other hand, layering process and rapid heating and cooling cause accumulation of a large amount of residual stresses that can adversely affect the mechanical properties. A heat treatment for removing residual stresses through recovery and recrystallization process caused complicated changes in mechanical properties induced by phase transformation, precipitate and homogenization of the microstructures. The purpose of this review was to compare the manufacturing methods of Co-Cr alloys and to investigate the characteristics of Co-Cr alloys fabricated by selective laser melting. (J Korean Acad Prosthodont 2021;59:248-60)

Keywords
Cobalt-chromium; Heat treatment; Mechanical property; Microstructure; Selective laser melting

Introduction
Cobalt-chromium (Co-Cr) alloys have emerged as promising alternatives to the precious metal alloys for the fabrication of various prostheses owing to their
low cost and satisfactory material properties, such as high strength, elastic modulus, corrosion resistance, and biocompatibility. Generally, metal frameworks of dental prostheses are fabricated using the casting technique; however, the complicated laboratory processes, porosities, and solidification shrinkage associated with this technique have gradually decreased the application of this method. With the development of computer-aided design and computer-aided manufacturing (CAD-CAM) system, CAD-CAM milling has emerged as an alternative method for the fabrication of dental prostheses. This method addresses the problems related to the complicated laboratory processes and processing time of the casting process; however, material wastage and the high wear and consumption rates of the milling tools limit the application of this method. The limitations of these conventional methods are overcome by the use of the additive manufacturing (AM) process, which has emerged as an effective method for the fabrication of dental prostheses. Among the several powder bed fusion (PBF) AM processes, selective laser melting (SLM) has been successfully applied for the fabrication of dental prostheses.\(^1\) SLM processes are characterized by a rapid local heating and cooling process, which improves the mechanical properties of the SLM Co-Cr alloys via the solid solution strengthening effect, secondary phase strengthening effect, grain refinement, and the reduction of porosities.\(^2\) However, a post-treatment (heat treatment) process is required to reduce the accumulated residual stresses caused by the layering process and the rapid heating and cooling in the SLM Co-Cr alloys. Heat treatment reduces the residual stresses of the SLM Co-Cr alloys not only by a recovery and recrystallization process but also by a phase transformation, which changes the behavior of the precipitates and the homogenization of the microstructures. Consequently, heat treatment leads to complex changes in the mechanical properties of the SLM Co-Cr alloys. In this literature review, various fabrication processes of Co-Cr alloys are compared, and the general characteristics of Co-Cr alloys and SLM Co-Cr alloys, and the effects of heat treatment on the SLM Co-Cr alloys were investigated.

**Manufacturing method of dental Co-Cr alloys**

Traditionally, the metal frameworks of prostheses are fabricated by a casting technique which is based on the lost-wax technique.\(^2\) These prostheses were commonly fabricated using precious metal alloys such as gold alloys; however, these alloys were gradually replaced with cheaper, nonprecious metal alloys such as nickel Ni-Cr and Co-Cr alloys, which exhibit relatively superior mechanical properties.\(^3\) Although Ni-Cr alloys were often used in the past, the risk of Ni-induced allergic reactions and the carcinogenic effect of beryllium have restricted their further application.\(^4,5\) Consequently, Co-Cr alloys have emerged as promising nonprecious alloys for the fabrication of prostheses for patients allergic to Ni.\(^6\) In addition, compared to the precious alloys, Co-Cr alloys are relatively inexpensive and exhibit favorable material properties such as high strength, corrosion resistance, tarnish resistance, and biocompatibility.\(^3,4,6\)

The fabrication of Co-Cr alloys via the casting technique is a relatively complex procedure involving waxing, investing, burnout, casting, finishing, and polishing. In addition, defects such as porosity and shrinkage are induced in the alloy during the casting process.\(^7\) Furthermore, among dental alloys that are commonly used, except the titanium alloy, Co-Cr alloys have the highest range of melting points. This poses restrictions on the possible extent of their manipulation in a dental laboratory. Moreover, owing to their high strength and low ductility, it is difficult to finish and polish Co-Cr alloys.\(^2,8\)

Recently, efforts have been devoted to fabricate Co-Cr alloys using the CAD-CAM system with the aim of reducing the processing time, while also reducing laboratory processes-related errors. The CAD-CAM system includes a subtractive manufacturing unit such as, milling and AM unit such as, PBF.\(^5,7,9,10\) During the CAD-CAM milling process, blocks of the materials are cut using a rotary
The generation of defects and porosities during this process may be minimized by employing Co-Cr alloy blocks fabricated via a standardized industrial process. However, owing to the rigidity of the ‘solid’ Co-Cr alloy blocks, it is difficult to achieve the CAD-CAM milling of these blocks; therefore, the dry milling of ‘soft’ Co-Cr alloy blocks has been introduced as an alternative method. During dry milling, a block containing a mixture of a combustible organic binder and the alloy powders are milled, and the milled structure is sintered in a high temperature furnace under argon atmosphere at approximately 1300°C until the block is fully densified. Dry milling enables the fabrication of dental restorations using a CAD-CAM device that is generally available in dental laboratories and reduces the processing time and cost. However, several factors limit the further application of subtractive manufacturing for the fabrication of dental prostheses such as high wastage of raw materials, short life time of milling tools owing to abrasive wears, increased abrasion on equipment, and high maintenance costs.

AM is an alternative method for the fabrication of dental prostheses and involves the use of powder- or liquid-based materials to fabricate solid structures. PBF is the most common AM method for the fabrication of metals in the field of dentistry and is largely divided into selective laser sintering (SLS) and SLM. SLS involves the incomplete sintering of powder particles within a specific area using the energy intensity per unit area of laser irradiation. However, during the partial sintering of the powders by lasers, voids are generated between the powders owing to theballing phenomenon, thus negatively affecting the strength and metal-ceramic bond strength of the fabricated structure. SLM is commonly used to fabricate metal frameworks because it shortens the time required for the manufacturing process, thus reducing the manufacturing labor and cost. SLM has attracted attention for the fabrication of desired parts owing to its minimal material use and waste and the ability to recycle the residual alloy powders. SLM involves the use of a high-energy laser such as a CO₂ laser or a fiber laser to melt the alloy powders. For the SLM fabrication of dental parts, first, CAD is employed to obtain the three-dimensional data of the complex shapes of dental prostheses, and divide the shape into several horizontal or vertical layers, after which the data are sent to the laser sintering machine. Subsequently, fine alloy powders, which are used as the raw materials, are applied to the powder bed. The alloy powders are intensively irradiated and heated using a high-energy laser. The alloy powders are rapidly melted and cooled locally to build up several layers, and this process is repeated until the designed structure in the CAD file is completely formed. SLM can completely melt alloy powders to fabricate a complicated, dense structure with almost no pores.

Microstructures and mechanical properties of cast Co-Cr alloys

Cast Co-Cr alloys, which are the most clinically used Co-Cr alloys and are used as a control group in experiments on SLM Co-Cr alloys, have the following characteristics. Cast Co-Cr alloys are composed of a face-centered cubic (FCC) lattice, γ phase and hexagonal close packed (HCP) lattice, and ε phase, and are thermodynamically stable at high and low temperatures. A slow FCC to HCP transformation in Co-Cr alloys leads to a metastable FCC phase at room temperature, which is related to the properties of Co-Cr alloys such as high strength, fatigue strength, and stress absorption. In addition, intermetallic compounds and carbides are usually precipitated at the grain boundaries (GBs) and interdendritic spaces of Co-Cr alloys. Although these precipitates contribute to the reinforcement mechanism, the inhomogeneous distribution, shapes, and sizes of the precipitates can reduce the ductility and fatigue strength of cast Co-Cr alloys. The cast Co-Cr alloys typically possess coarse grains; however, the strength of these alloys increases with an increase in the cooling rate and a decrease in the grain size.
be employed to analyze the microstructures of alloys, including the grain size and distribution, GB, and strain by misorientation. Fig. 1 shows the EBSD data of the cast Co-Cr and SLM Co-Cr alloys. As shown in the image, the cast Co-Cr alloys exhibit equiaxed coarse grains (Fig. 1A, B). The color differences of the grains in the image indicate a difference in the crystal orientation.

A major limitation of the cast Co-Cr alloys is their inherent porosities, which are formed due to solidification and shrinkage during the casting process. Cast Co-Cr alloys form typical dendritic structures owing to their low cooling and solidification rates and negative temperature gradient. During the formation of these dendritic structures, the interdendritic regions separate from the melt so that the resultant shrinkage in the solidified alloy is not from the melt, thus leading to the formation of interdendritic micropores (Fig. 2A). Generally, the porous parts of a metal framework undergo drastic changes in the cross-section thickness. These changes affect the tensile strength and ductility of the alloy, thus leading to unexpected fractures in the frameworks of removable partial dentures such as clasps. The generation of internal pores on the surface of the alloy after the adjustment of the prostheses can adversely affect the mechanical strength of the prostheses or reduce the bond strength at the metal-ceramic bonding site owing to the formation of pores at the metal-ceramic bonding site. In addition, the formation of a second phase, which is usually a molybdenum (Mo)-rich phase, is commonly observed in the interdendritic regions, as indicated by the relatively white area in Fig. 2A. The dispersion of the second phase increases the brittleness of the alloy by removing Mo from the solid solution, thus negatively affecting the mechanical properties of the alloy and increasing the vulnerability of the alloy to various types of corrosion such as crevice corrosion and pitting corrosion.

**Microstructures and mechanical properties of SLM Co-Cr alloys**

The SLM process is characterized by a rapid local heating and cooling of the material. During this process, a
high-energy laser beam is used to irradiate the alloy powders, which absorb the energy through bulk coupling or powder coupling. Consequently, high temperatures and cooling rates are generated within the molten pool. In a Co-Cr binary system, the FCC phase has low energy, $G$, whereas the HCP phase is thermodynamically stable at room temperature. However, in SLM Co-Cr alloys, most of the FCC phase is maintained at room temperature owing to the rapid cooling process. Consequently, the solid solution limit of the alloying elements increases, thus reducing the precipitate and dendritic segregation, while maintaining oversaturation. This leads to the solid solution strengthening effect and second phase strengthening effect. Therefore, SLM Co-Cr alloys exhibit higher strength, hardness, and ductility than the cast Co-Cr alloys owing to the higher FCC phase fractions and solid solution limits of SLM Co-Cr alloys.

The grain size of the SLM Co-Cr alloys is closely related to the nucleation rate ($I$) and is determined by the degree of undercooling ($\Delta T$). Owing to the larger $\Delta T$ produced by the SLM technique than that produced by the casting technique, SLM Co-Cr alloys possess a finer grain size and higher strength and hardness compared to cast Co-Cr alloys. For example, Santos et al. observed column grains with a mean size of 55 $\mu$m (length) and 22 $\mu$m (height) in SLM Co-Cr alloys. The improvement in the mechanical properties of SLM Co-Cr alloys owing to a reduced grain size can be expressed in terms of the Hall-Petch (H-P) relationship and can be calculated using Equation 1:

$$\Delta \sigma_s = kD^{-1/2},$$

where $D$ is the mean grain size, and $k$ is the slope of the H-P relationship line that measures the relative extent of the strength contribution. Grain refinement strengthening is an important factor of SLM that improves the mechanical properties of the alloys, enhancing not only the strength but also the ductility and toughness of the alloys. GBs act as a hindrance to crack propagation. With a decrease in the grain size, the GB area increases, thus enhancing the effect of grain refinement strengthening and further improving the mechanical properties of the alloys. In addition, because the primary dendrite spacing in the fine microstructures of the SLM Co-Cr alloys are significantly smaller (2.5 $\mu$m) compared to that of the cast Co-Cr alloys, grain refinement strengthening reduces the slip length in the alloy, thus improving the strength of the alloy (Fig. 2B). As shown in Fig. 1, the GB area of SLM Co-Cr alloy (Fig. 1D, E) is larger than that of cast Co-Cr alloy (Fig. 1A, B).

There are several SLM factors that affect the properties of the final product including the choice of powders, building direction or orientation, and processing parameters. The building direction, which is the acute angle between the longitudinal axis of a sample and the building platform, affects the microstructures, texture, and residual stresses of the final product (Fig. 3A). Thermal fluctuations that occur during the SLM manufacturing process have the most significant effect on the building direction of the formed grains, which generally grow from the building platform with lower temperatures to

---

**Fig. 2.** Backscattered electron (BSE) images of polished Co-Cr alloys. Black arrows indicate microporosities. (A) Cast Co-Cr, (B) SLM Co-Cr.
the upper surface with higher temperatures. In addition, the building direction affects the arrangement of precipitates. Several studies have reported on molten pool boundaries (MPBs), which are characteristically observed in SLM alloys and significantly affect their mechanical properties (Fig. 3B). The number and angle of MPBs vary depending on the building direction and can therefore affect the anisotropy of the mechanical properties. As the number of layers at 0° and 45° are higher than those at 90°, a larger number of MPBs are formed. With an increase in the number of layers, the likelihood of defects or porosity also increases, thus resulting in a mechanical anisotropy. In addition, the ductile deformation of SLM alloys results from grain slipping and MPBs. Owing to the relatively weak bond strength of MPBs compared to that of GBs, slipping occurs preferentially along the MPBs. Kajima et al. compared the anisotropy of the fatigue strength after setting the building direction of a clasp to 0°, 45°, and 90°. They found that the SLM Co-Cr alloys exhibited higher yield and tensile strengths compared to the cast Co-Cr alloys for all directions and attributed this to the microstructures of the SLM Co-Cr alloys, which were formed by rapid cooling. However, they also observed a mechanical anisotropy in the SLM Co-Cr alloys and a lower fatigue strength when the clasps were set parallel to the building direction. In addition, they reported that because fractures tend to be parallel to the MPBs, cracks would easily advance along the MPBs.

During the SLM process, unique microstructures similar to ‘fish scales’ appear based on the molten pool and layer stacks of each laser path. Under high magnification, fine microstructures are observed as “fish scales,” and numerous columnar grains growing perpendicular to the circular arch-shaped MPBs are also observed. Additionally, large interlocked elongated grains with different directions are observed in the dendritic structures, and the ‘weld line’ and the three-dimensional texture of these grains, which is similar to those of common weaved fabrics can improve the damage tolerance of the alloy. The columnar-cellular growth of grains is associated with the direction of heat flow. Fine cellular- or columnar-dendritic structures act as obstacles to dislocation motions and may increase the tensile strength of the SLM Co-Cr alloys.

The microstructures and mechanical properties of SLM alloys are affected by the SLM process parameters such as the laser power, scanning velocity, hatch spacing, laser beam size, and layer thickness, which are associated with the melting energy and the penetration depth of energy in the alloy powders, which consequently affect the density of the molten alloy. Takaichi et al. reported that the porosities of SLM Co-Cr alloys could be reduced by optimizing the process parameters, which include the laser power, hatch spacing, scanning speed, and layer thickness. The laser energy density (LED) of the SLM pro-

---

**Fig. 3.** Schematic diagram of the layers fabricated by SLM. (A) Anisotropy based on the different building direction, (B) Molten pool boundary (MPB).
cess can be calculated using four major variables: laser power $P$ (W), scanning velocity $V$ (mm/s), hatch spacing $H$ (mm), and layer thickness $D$ (mm), as shown in Equation 2.  

$$\text{LED} = \frac{P}{VHD} \text{ (J/mm}^3\text{)}$$  

(2)

Takaichi et al. examined the correlation between the LED and porosities in SLM Co-Cr alloys and reported that porous structures develop at a LED of $<$150 J/mm$^3$. In addition, Qian et al. reported that the tensile strength increases at an LED of 118 J/mm$^3$ than that at an LED of 63 J/mm$^3$. Additionally, Tonelli et al. reported a lack of fusion and presence of large amounts of internal porosities at an LED of $<$100 J/mm$^3$ and keyhole collapse at an LED of $>$200 J/mm$^3$. Furthermore, studies have reported that optimum mechanical properties are obtained at LED values of 150 - 200 J/mm$^3$. Owing to the negative effect of high porosities on the mechanical properties of the SLM Co-Cr alloys, achieving a near-full density at an adequate LED is important.

As shown in the backscattered electron (BSE) images, a small amount of porosity can be observed in the SLM Co-Cr alloys compared to the cast Co-Cr alloys (Fig. 2B).

As the kernel average misorientation (KAM) value principally indicates the microscopic lattice distortion, the distribution of residual stresses can be indirectly measured using the KAM method. As shown in Fig. 1C, the KAM value of the SLM Co-Cr alloy (Fig. 1F) is higher than that of the cast Co-Cr alloy, indicating a high distribution of residual stresses in the SLM Co-Cr alloy. The accumulation of such a large amount of residual stresses could be attributed to the rapid heating and cooling of the SLM, which may adversely affect the mechanical properties. In addition, owing to the layer-by-layer additive process of SLM, a significant temperature difference exists between the previously printed layer and the newly deposited layer, and this difference is one of the factors that cause residual stresses. The temperature gradient mechanism and cool-down mechanism based on the expansion behaviors of a material during heating or cooling can be used to explain the generation of residual stresses (Fig. 4). For example, under the assumption that the materials are sufficiently melted and metallurgical connections occur between the layers, strain accommodation occurs in the material and residual stresses are formed. In a laser-based process, a very high cooling rate of the order of $10^3 - 10^8$ K/s forms a steep thermal gradient, resulting in high residual stresses, which are close to the yield strength of the material. When the residual stresses exceed the yield strength, heat is retained in the material in the form of residual stresses even after the complete removal of heat. Consequently, this leads to distortions such as bending, warping, pore formation, cracking, delamination, and plastic deformation, thus compromising the mechanical properties. Furthermore, premature fractures can occur even under

**Fig. 4.** Temperature gradient mechanism suggested by Simson et al. (A) Residual stress based on heating and cooling ($\varepsilon_{th}$: thermal elongation, $\varepsilon_{pl}$: plastic elongation, $\sigma_{\text{comp}}$: compressive stress, $\sigma_{\text{tens}}$: tensile stress), (B) Residual stress at the solid layer connection.
low cycle loads owing to the reduced fatigue strength. Consequently, these problems restrict the immediate application of the components after their fabrication. Thus, methods such as the preheating of powders and the baseplate, island scanning strategy, re-scanning, and heat treatment have been introduced to reduce the residual stress of SLM Co-Cr alloys. Among these methods, heat treatment is the most economic and effective method for removing residual stresses. Heat treatment enables the complete removal of the residual stresses in the alloy, while ensuring the formation of homogeneous microstructures, thus improving the mechanical properties of the alloys.

The microstructural differences that affect the mechanical properties of cast Co-Cr alloys and SLM Co-Cr alloys are summarized in Table 1.

**Effect of heat treatment on the microstructures and mechanical properties of SLM Co-Cr alloys**

Depending on the heat treatment conditions, alloys undergo recovery, recrystallization, and growth, and consequently, grain changes. During deformation, energy is primarily stored in the material in the form of dislocations, which are considered as the crystallographic defect sites. During the recovery process, grains reduce stored energy by removing defects such as dislocations within the crystal structure or rearranging dislocations. The internal strain energy is removed as atoms move from the higher stress areas to lower stress areas. While heat treatment must be performed at sufficiently high temperatures that permit atomic mobility, it should be performed for a short period, in order to prevent undesired recrystallization and grain growth, which are associated with strength loss. Kajima et al. and Takaichi et al. reported that the recovery process of SLM Co-Cr alloys occurs after heat treatment at 1050°C or less for 6 h, and recrystallization occurred after heat treatment at 1150°C for 6 h. In contrast, some other studies have reported that recrystallization occur in SLM Co-Cr alloy after heat treatment at 1150°C for 1 h or at 1220°C. This indicates that an appropriate temperature is more important for the reduction of the residual stresses by the recrystallization process than the heat treatment time. Moreover, as the strength and hardness decrease with an increase in grain size, further studies are needed to control the degree of recrystallization and growth.

Heat treatment also affects the FCC → HCP phase transformation, and the changes in the mechanical properties of SLM Co-Cr alloys primarily depend on the changes in the phase fraction of the brittle HCP phase. With a reduction in the HCP phase after heat treatment, the strength and hardness decreases, thus increasing the ductility. Additionally, heat treatment affects the behavior of the precipitate and the homogenization of the microstructures. As precipitates exhibit a secondary phase strengthening effect, a decrease in the precipitates after heat treatment leads to a decrease in the strength and hardness of the SLM Co-Cr alloy. In addition, the homogenization of the microstructures and the reduction of mechanically weak MPBs after heat

| Table 1. Microstructural differences influencing mechanical properties between cast Co-Cr alloys and SLM Co-Cr alloys |
|-------------------------------|-----------------|---------------------|
| **Phase fraction** | Cast Co-Cr | SLM Co-Cr (as-built) | References |
| **Precipitate** | FCC + HCP, dual phase | High FCC, low HCP | 2,8,9,15-17 |
| **Grain size** | Coarse | Low content, small size | 4,8,9,17,19 |
| **Porosity** | High | Fine | 2,17,19,27 |
| **Residual stress** | Low | Low | 4,8,16,18,30,39 |

https://doi.org/10.4047/jkap.2021.59.2.248
treatment reduces the anisotropy of mechanical properties and increases the fatigue strength. \cite{33,34,36,43} Therefore, a heat treatment that can properly control the required mechanical properties should be considered. The changes in the mechanical properties of SLM Co-Cr alloys based on heat treatment are summarized in Table 2.

In summary, during heat treatment, residual stresses are removed through the recovery and recrystallization processes, and the heat treatment temperature has a more significant effect on the efficiency of the heat treatment than the heat treatment time. In addition, heat treatment affects the FCC → HCP phase transformation, the behavior of the precipitates, and the homogenization of microstructures. The mechanical properties of SLM Co-Cr alloys are closely related to the FCC → HCP phase transformation, and the strength and hardness of the alloy increase with an increase in the phase fraction of the HCP phases, whereas the ductility decreases with an increase in the HCP phase. In addition, the strength and hardness decrease with a reduction in the amount of precipitates. Furthermore, the homogenization of the microstructures including the MPBs reduces the anisotropy of mechanical properties and increases the fatigue strength of SLM Co-Cr alloys.

**Conclusion**

In summary, Co-Cr alloys are commonly used to fabricate dental prostheses owing to their cost-effectiveness and excellent mechanical properties. SLM overcomes the limitations of existing manufacturing methods for the fabrication of dental prostheses using Co-Cr alloys. The rapid local heating and cooling in the SLM process results in the formation of fine microstructures, grain

**Table 2. Mechanical properties of Co-Cr alloys fabricated by SLM and subsequently heat treated**

| Reference      | Heat treatment | Building orientation | Ultimate tensile strength (MPa) | 0.2% Yield strength (MPa) | Vickers Hardness (HV) | Elongation (%) |
|----------------|----------------|----------------------|---------------------------------|--------------------------|----------------------|---------------|
| Kajima et al. 2018 | As-built       | Longitudinal         | 1173 ± 25                       | 839 ± 32                 | 477 ± 9              | 12.3 ± 2.8    |
|                 | 750°C, 1h*     |                      | 1097 ± 11                       | 953 ± 16                 | 498 ± 2              | 3 ± 0.5       |
|                 | 900°C, 1h      |                      | 1071 ± 14                       | 793 ± 25                 | 495 ± 10             | 5.8 ± 0.9     |
|                 | 1050°C, 1h     |                      | 1075 ± 11                       | 738 ± 4                  | 428 ± 9              | 11.4 ± 0.7    |
|                 | 1150°C, 1h     |                      | 1007 ± 32                       | 614 ± 11                 | 365 ± 11             | 16.3 ± 1.4    |
| Wei et al. 2020 | As-built       | Longitudinal         | 1142 ± 21                       | 672 ± 24                 | -                    | 8.0 ± 0.2     |
|                 | 750°C, 1h      |                      | 1279 ± 16                       | 1019 ± 33                | -                    | 4.1 ± 0.3     |
|                 | 850°C, 1h      |                      | 1463 ± 19                       | 1258 ± 27                | -                    | 1.7 ± 0.2     |
|                 | 950°C, 1h      |                      | 1228 ± 21                       | 973 ± 14                 | -                    | 1.6 ± 0.2     |
|                 | 1050°C, 1h     |                      | 1189 ± 20                       | 754 ± 19                 | -                    | 10.1 ± 0.3    |
|                 | 1150°C, 1h     |                      | 920 ± 14                        | 516 ± 18                 | -                    | 12.2 ± 0.4    |
| Zhou et al. 2020 | As-built       | Longitudinal         | 1113 ± 25                       | 795 ± 11                 | -                    | 9.8 ± 2.8     |
|                 | As-built PF**-treated |                  | 1325 ± 11                       | 1080 ± 14                | -                    | 3.2 ± 0.9     |
|                 | 850°C, 1h PF-treated |            | 1366 ± 32                       | 1056 ± 16                | 543.7                | 4.33 ± 2.59   |
|                 | 950°C, 1h PF-treated |            | 1410 ± 11                       | 914 ± 25                 | 557.9                | 6.05 ± 0.53   |
|                 | 1050°C, 1h PF-treated |          | 1458 ± 14                       | 956 ± 11                 | 537.6                | 7.73 ± 0.97   |
|                 | 1150°C, 1h PF-treated |          | 1296 ± 11                       | 816 ± 4                  | 402.6                | 12.20 ± 0.77  |

* h: hour

** PF: Porcelain firing schedules including degassing and oxidation, opaque porcelain, body porcelain, enamel porcelain
refinement, and the reduction of the porosities in the alloys. Thus, SLM Co-Cr alloys exhibit better mechanical properties such as strength, hardness, and ductility compared to cast Co-Cr alloys. However, the layering process and rapid heating and cooling cause the formation of inhomogeneous microstructures and the accumulation of a large amount of residual stresses, which negatively affect the mechanical properties of SLM Co-Cr alloys. Generally, heat treatment is employed to homogenize the microstructures and remove residual stresses in SLM Co-Cr alloys through recovery and recrystallization processes. Heat treatment affects the phase transformation and the behavior of the precipitates, resulting in complex changes in the mechanical properties including strength, hardness, and ductility.

Acknowledgments

The authors would like to acknowledge KITECH (Korea Institute of Industrial Technology) for technical assistance.

References

1. Tulga A. Effect of annealing procedure on the bonding of ceramic to cobalt-chromium alloys fabricated by rapid prototyping. J Prosthet Dent 2018;119:643-9.
2. Kim HR, Jang SH, Kim YK, Son JS, Min BK, Kim KH. Microstructures and mechanical properties of Co-Cr dental alloys fabricated by three CAD/CAM-based processing techniques. Materials (Basel) 2016;9:596.
3. Suleiman SH, Vult von Steyern P. Fracture strength of porcelain fused to metal crowns made of cast, milled or laser-sintered cobalt-chromium. Acta Odontol Scand 2013;71:1280-9.
4. Al Jabbari YS. Physico-mechanical properties and prostodontic applications of Co-Cr dental alloys: a review of the literature. J Adv Prosthodont 2014;6:138-45.
5. Stawarczyk B, Eichberger M, Hoffmann R, Noack F, Schweiger J, Edelhoff D. A novel CAD/CAM base metal compared to conventional CoCrMo alloys: an in-vitro study of the long-term metal-ceramic bond strength. Oral Health Dent Manag 2014;13:446-52.
6. Hedberg YS, Qian B, Shen Z, Virtanen S, Wallinder IO. In vitro biocompatibility of CoCrMo dental alloys fabricated by selective laser melting. Dent Mater 2014;30:525-34.
7. Konieczny B, Szczesio-Wlodarczyk A, Sokolowski J, Bociong K. Challenges of Co-Cr alloy additive manufacturing methods in dentistry - the current state of knowledge (systematic review). Materials (Basel) 2020;13:3524.
8. Takaichi A, Suyalatu, Nakamoto T, Joko N, Nomura N, Tsutsumi Y. Microstructures and mechanical properties of Co-29Cr-6Mo alloy fabricated by selective laser melting process for dental applications. J Mech Behav Biomed Mater 2013;21:67-76.
9. Al Jabbari YS, Koutsoukis T, Barmpagadaki X, Zinelis S. Metallurgical and interfacial characterization of PFM Co-Cr dental alloys fabricated via casting, milling or selective laser melting. Dent Mater 2014;30:e79-88.
10. Koutsoukis T, Zinelis S, Eliades G, Al-Wazzan K, Rifaiy MA, Al Jabbari YS. Selective laser melting technique of Co-Cr dental alloys: a review of structure and properties and comparative analysis with other available techniques. J Prosthodont 2015;24:303-12.
11. Lee DH, Lee BJ, Kim SH, Lee KB. Shear bond strength of porcelain to a new millable alloy and a conventional castable alloy. J Prosthet Dent 2015;113:329-35.
12. Revilla-León M, Sadeghpour M, Özcan M. A review of the applications of additive manufacturing technologies used to fabricate metals in implant dentistry. J Prosthodont 2020;29:579–593.
13. Bae EJ, Kim JH, Kim WC, Kim HY. Bond and fracture strength of metal-ceramic restorations formed by selective laser sintering. J Adv Prosthodont 2014;6:266-71.
14. Ayyildiz S, Soylu EH, Ide S, Kilic S, Sipahi C, Piskin B. Annealing of Co-Cr dental alloy: effects on nanostructure and Rockwell hardness. J Adv Prosthodont 2013;5:471-8.
15. Zhou Y, Sun Q, Dong X, Li N, Shen ZJ, Zhong Y. Microstructure evolution and mechanical properties improvement of selective laser melted Co-Cr bio-

https://doi.org/10.4047/jkap.2021.59.2.248
medical alloys during subsequent heat treatments. J Alloys Compd 2020;840:155664.
16. Giacchi JV, Morando CN, Fornaro O, Palacio HA. Microstructural characterization of as-cast biocompatible Co-Cr-Mo alloys. Mater Charact 2011;62:53-61.
17. Kaiser R, Williamson K, O’Brien C, Ramirez-Garcia S, Browne DJ. The influence of cooling conditions on grain size, secondary phase precipitates and mechanical properties of biomedical alloy specimens produced by investment casting. J Mech Behav Biomed Mater 2013;24:53-63.
18. Krawczyk MB, Krolikowski MA, Grochala D, Powałka B, Figiel P, Wojciechowski S. Evaluation of surface topography after face turning of CoCr alloys fabricated by casting and selective laser melting. Materials (Basel) 2020;13:2448.
19. Zhou Y, Li N, Yan J, Zeng Q. Comparative analysis of the microstructures and mechanical properties of Co-Cr dental alloys fabricated by different methods. J Prosthet Dent 2018;120:617-23.
20. Lewis AJ. Radiographic evaluation of porosities in removable partial denture castings. J Prosthet Dent 1978;39:278-281.
21. Fischer P, Romano V, Weber HP, Karapatis NP, Boillat E, Glardon R. Sintering of commercially pure titanium powder with a Nd:YAG laser source. Acta Materialia 2003;51:1651-62.
22. Liu Y, Yang Y, Mai S, Wang D, Song C. Investigation into spatter behavior during selective laser melting of AISI 316L stainless steel powder. Mater Des 2015;87:797-806.
23. Xin XZ, Chen J, Xiang N, Wei B. Surface properties and corrosion behavior of Co-Cr alloy fabricated with selective laser melting technique. Cell Biochem Biophys 2013;67:983-90.
24. Santos CD, Habibe AF, Simba BG, Lins JFC, Freitas BXd, Nunes CA. CoCrMo-base alloys for dental applications obtained by selective laser melting (SLM) and CAD/CAM milling. Mat Res 2020;23. https://doi.org/10.1590/1980-5373-mr-2019-0599.
25. Song B, Dong S, Deng S, Liao H, Coddet C. Microstructure and tensile properties of iron parts fabricated by selective laser melting. Opt Laser Technol 2014;56:451-60.
26. Sing SL, Huang S, Yeong WY. Effect of solution heat treatment on microstructural and mechanical properties of laser powder bed fusion produced cobalt-28chromium-6molybdenum. Mater Sci Eng A 2020;769:138511.
27. Meacock CG, Vilar R. Structure and properties of a biomedical Co-Cr-Mo alloy produced by laser powder. J Laser Appl 2009;21:88-95.
28. Shifeng W, Shuai L, Qingsong W, Yan C, Sheng Z, Yusheng S. Effect of molten pool boundaries on the mechanical properties of selective laser melting parts. J Mater Process Technol 2014;214:2660-7.
29. Barucca G, Santecchia E, Majni E, Girardin E, Bassoli E, Denti L. Structural characterization of biomedical Co-Cr-Mo components produced by direct metal laser sintering. Mater Sci Eng C Mater Biol Appl 2015;48:263-9.
30. Qian B, Saedik I, Kvetkova L, Lofaj F, Xiao C, Shen Z. Defects-tolerant Co-Cr-Mo dental alloys prepared by selective laser melting. Dent Mater 2015;31:1435-44.
31. Kajima Y, Takaichi A, Nakamoto T, Kimura T, Yogo Y, Ashida M, Doi H, Nomura M, Takahashi H, Hanawa T, Wakabayashi N. Fatigue strength of Co-Cr-Mo alloy clasps prepared by selective laser melting. J Mech Behav Biomed Mater 2016;59:446-58.
32. Takaichi A, Kajima Y, Kittikundecha N, Htay HL, Wat Cho HH, Hanawa T. Effect of heat treatment on the anisotropic microstructural and mechanical properties of Co-Cr-Mo alloys produced by selective laser melting. J Mech Behav Biomed Mater 2020;102:103496.
33. Seki E, Kajima Y, Takaichi A, Kittikundecha N, Cho HHW, Htay HL. Effect of heat treatment on the anisotropic microstructural and mechanical properties of Co-Cr-Mo alloys fabricated by selective laser melting. Mater Lett 2019;245:53-6.
34. Girardin E, Barucca G, Mengucci P, Fiori F, Bassoli E, Gatto A. Biomedical Co-Cr-Mo components produced by direct metal laser sintering. Mater Today 2016;3:889-97.
35. Zhang M, Yang Y, Song C, Bai Y, Xiao Z. An investigation into the aging behavior of CoCrMo alloys fabricated by selective laser melting. J Alloys Compd 2018;750:878-86.
36. Wei W, Zhou Y, Liu W, Li N, Yan J, Li H. Microstructural characterization, mechanical properties, and
corrosion resistance of dental Co-Cr-Mo-W alloys manufactured by selective laser melting. J Mater Eng Perform 2018;27:5312-20.

37. Averyanova M, Bertrand P, Verquin B. Manufacture of Co-Cr dental crowns and bridges by selective laser melting technology. Virtual Phys Prototyp 2011; 6:179-85.

38. Bartlett JL, Li X. An overview of residual stresses in metal powder bed fusion. Addit Manuf 2019;27:131-49.

39. Tonelli L, Fortunato A, Ceschini L. CoCr alloy processed by selective laser melting (SLM): effect of laser energy density on microstructure, surface morphology, and hardness. J Manuf Process 2020;52:106-19.

40. Simson T, Emmel A, Dwars A, Böhm J. Residual stress measurements on AISI 316L samples manufactured by selective laser melting. Addit Manuf 2017;17:183-9.

41. Mugwagwa L, Dimitrov D, Matope S, Yadroitsev I. Influence of process parameters on residual stress related distortions in selective laser melting. Procedia Manuf 2018; 21:92-99.

42. Mengucci P, Barucca G, Gatto A, Bassoli E, Dentil L, Fiori F. Effects of thermal treatments on microstructure and mechanical properties of a Co-Cr-Mo-W biomedical alloy produced by laser sintering. J Mech Behav Biomed Mater 2016;60:106-17.

43. Doherty RD, Hughes DA, Humphreys FJ, Jonas JJ, Juul Jensen D, Kassner ME, King WE, McNelley TR, McQueen HJ, Rollett AD. Current issues in recrystallization: a review. Mater Sci Eng 1997;238:219-274.

44. Sames WJ, List FA, Pannala S, Dehoff RR, Babu SS. The metallurgy and processing science of metal additive manufacturing. Int Mater Rev 2016;61:315-60.

45. Kajima Y, Takaichi A, Kittikundecha N, Nakamoto T, Kimura T, Nomura N, Kawasaki T, Takahashi H, Wakabayashi N. Effect of heat-treatment temperature on microstructures and mechanical properties of Co-Cr-Mo alloys fabricated by selective laser melting. Mater Sci Eng A 2018;726:21-31.

46. Sing SL, Huang S, Yeong WY. Effect of solution heat treatment on microstructure and mechanical properties of laser powder bed fusion produced cobalt-28chromium-6molybdenum. Mater Sci Eng A

2020;769:138511.

47. Kittikundecha N, Kajima Y, Takaichi A, Wai Cho HH, Htat HL, Doi H. Fatigue properties of removable partial denture clasps fabricated by selective laser melting followed by heat treatment. J Mech Behav Biomed Mater 2019;98:79-89.

48. Wei W, Zhou Y, Sun Q, Li N, Yan J, Li H et al. Microstructures and mechanical properties of dental Co-Cr-Mo-W alloys fabricated by selective laser melting at different subsequent heat treatment temperatures. Metall Mater Trans A 2020;51:3205-14.

49. Kittikundecha N, Kajima Y, Takaichi A, Wai Cho HH, Htat HL, Doi H. Fatigue properties of removable partial denture clasps fabricated by selective laser melting followed by heat treatment. J Mech Behav Biomed Mater 2019;98:79-89.

https://doi.org/10.4047/jkap.2021.59.2.248
선택적 레이저 용융 방법으로 제작한 치과용 코발트 크롬 합금에 대한 문헌고찰

강현구*
강릉원주대학교 치과대학 치과보철학교실 및 구강과학연구소

코발트-크롬 합금은 다양한 치과보철물 제작에 이용되고 있고, 다른 합금에 비해 저렴한 가격과 우수한 기계적 특성이 장점이다. 최근, 기존 제작 방식의 단점을 극복하기 위해 적층제조 방식인 선택적 레이저 용융 방법이 보철물 제작에 이용되고 있다. 선택적 레이저 용융 방법의 공정 중 급속 가열과 냉각 과정은 제작한 합금의 미세구조와 결정립을 미세화하고, 기포를 감소시키기 위해 제작 방식에 의한 합금에 비해 기계적 특성을 향상시킨다. 반면, 적층과 급속 가열 및 냉각은 다량의 잔류응력 축적을 초래하는데, 추후 기계적 특성에 악영향을 미칠 수 있다. 따라서, 잔류응력을 제거하기 위해 주로 열처리를 시행하고, 회복과 재결정화에 의한 잔류응력의 감소뿐만 아니라 상변태, 석출물 및 미세구조의 군집화가 동반되어 기계적 특성의 복잡한 변화가 나타난다. 본 문헌고찰에서 코발트-크롬 합금의 제작 방식 비교 및 선택적 레이저 용융 방법으로 제작된 합금의 특징에 대해 알아보고자 한다. (대한치과보철학회지 2021;59:248-60)

주요단어
코발트-크롬; 열처리; 기계적 특성; 미세구조; 선택적 레이저 용융

교신저자
강현구
25457 강원도 강릉시 죽헌길 7
강릉원주대학교 치과대학
치과보철학교실 및 구강과학연구소
033-640-3154
prostho9@naver.com

원고접수일 2020년 11월 12일
원고최종수정일 2021년 1월 7일
원고체택일 2021년 1월 20일

© 2021 대한치과보철학회
이 글은 크리에이티브 커먼즈 라이선스에 따라 이용하실 수 있습니다.