MECHANICAL PROPERTIES OF DENTAL Co-Cr ALLOYS FABRICATED VIA CASTING AND SELECTIVE LASER MELTING

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Abstract: *The aim of the present paper is to investigate the mechanical properties (hardness and tensile strength) of dental Co-Cr alloys fabricated via casting and selective laser melting (SLM). Two groups of metallic specimens (four-part dental bridges and standard tensile test specimens) made of Co–Cr dental alloys were produced by lost-wax casting and SLM processes. Vickers hardness distribution along the depth of the dental bridges as well as the Rockwell hardness and tensile strength of the samples were studied out. The hardness of Co–Cr dental alloys are dependent on the manufacturing technique employed. It was established that the average Vickers hardness of the samples, produced by SLM, was higher than that of the cast samples 382 HV and 335 HV respectively. The nearly even hardness distribution in the bridges, produced by SLM, and fluctuations of the hardness values along the depth of the cast bridges were observed. The Rockwell measurements confirmed the higher hardness of the SLM samples – 39 HRC in comparison with that of the cast ones – 33 HRC. The tensile strength is in good agreement with the hardness values. Due to the unique microstructure, the yield strength and tensile strength for the SLM samples were higher than those of the as-cast alloy.*

Keywords: DENTAL Co-Cr ALLOYS, CASTING, SELECTIVE LASER MELTING, HARDNESS, TENSILE STRENGTH

1. Introduction

Cobalt–chromium based alloys have been widely used in various orthopedic implants as well as for manufacturing of metal framework of fixed dental prosthesis because of their excellent mechanical properties, high corrosion and wear resistance, and good biocompatibility. The chemical composition of dental Co-Cr alloys consists of 53–67% Co, 25–32% Cr, 2–6% Mo and small quantities of W, Si, Al and others [1]. Chromium, molybdenum and tungsten are added for strengthening of the solid solution. Due to the relatively large amount of Cr, dense passive layer of $Cr_2O_3$with 1–4 nm thickness is formed on the surface of the details, determining the high corrosion resistance [2,3].

In proper alloying the microstructure of the dental alloys is composed mainly of γ-phase and carbides of the $M_{23}C_6$ type [3]. The high temperature γ-phase possess face centered cubic (FCC) lattice, while the room temperature ε-phase has hexagonal close packed (HCP) lattice [2-5]. The γ-phase determines ductility, while the-ε phase enhances the corrosion and wear resistance [7]. So, the properties of the Co-Cr dental alloys depend on the γ-ε ratio and the type, quantity and distribution of the carbide phase in the microstructure.

Most of the dental constructions are manufactured by lost-wax casting process which consists of many manual operations, leading to low accuracy and satisfactory quality. The new process of Selective Laser Melting (SLM) offers opportunity for overcoming the disadvantages of the casting process. In this technology layers of metal powder are fused into a 3D model by adopting a computer-directed laser [7-12]. The advantages of SLM over the traditional methods include production of personalized complex objects; manufacturing of parts with dense structure and predetermined surface roughness; controllable, easy and relatively quick process [7,13].

The SLM process characterizes with high heating and cooling rates of the melted layer as well as heating and solid state phase transformations in the underneath layers, which determine microstructure and properties quite different than that of the cast details.

Meacock et al. [14] reported that the microstructure of biomedical Co-Cr-Mo alloy, produced by laser powder microdeposition, is homogeneous comprised of fine cellular dendrites. The average hardness was 460 HV0.2, which is higher than the values obtained by the other fabrication process. Barucca et al. [15] investigated Co-Cr-Mo parts, produced by direct metal laser sintering. They established that microstructure consists of γ and ε phases. The ε phase is distributed as network of thin lamellae inside the γ-phase. The higher hardness is attributed to the presence of the ε-lamellae grown on the $\{111\}_\gamma$ planes that restricts the dislocations movement in the γ-phase. Yanjin Lu et al. [16] investigated the microstructure, hardness, mechanical properties, electrochemical behaviour and metal release of Co-Cr-W alloy fabricated by SLM in two different scanning strategies – line and island. They established the coexistence of the γ- and ε-phases in the microstructure and nearly the same hardness – 570 HV for line-formed alloy and 564 HV for island-formed. Their research show that the results of tensile, hardness, density, electrochemical and metal release tests are independent of the scanning strategy and the yield strength of both samples meet the ISO 22764:2006 standard for dental restorations. Wen Shifeng et al. [17] investigated the influence of the samples orientation during SLM manufacturing process on the tensile strength. They established that the SLM specimens made along the vertical direction have higher tensile strength and elongation than those made along the horizontal direction, indicating significant anisotropic feature of SLM parts. According to them the molten pool boundaries have a significant impact and are the main reason for anisotropy and low ductility of SLM parts.

SLM is comparatively new process and the data about the microstructure and mechanical properties of the constructions, manufactured using t, are relatively scarce. The purpose of this study was to fabricate a Co–Cr alloy using the SLM process and casting process, as well as to investigate the microstructure, and mechanical properties.

2. Materials and methods

Materials and samples preparation

In order to understand the relationship between the process, microstructure, surface hardness and mechanical properties, tensile tests with specimens made of Co-Cr alloy were carried out. Two groups of samples – four-part dental bridges (Fig. 1) and tensile test specimens were prepared by lost-wax casting and SLM using Co–Cr dental alloys.

In order to obtain samples with sufficiently good repeatability at first a base model of 4-part dental bridge was made. It was used for manufacturing of silicone mold for production of wax models

Fig. 1. *SLM of dental bridges*

and for generating of virtual 3D model. A silicone mold for manufacturing of wax models for the cast tensile test specimens was also fabricated, while the 3D model was created with SolidWorks software. The cast samples – bridges and tensile test specimens were produced by centrifugal casting of Co-Cr alloy “Biosil” with chemical composition, given by the producer (Table 1).

Table 1.
Chemical composition of the alloys used.

Alloy	Chemical composition, mass %							
	Co	Cr	Mo	Si	Mn	C	Fe	Ni
ASTM F75	Bal.	27-30	5-7	<1	<1	<0.35	<0.75	<0.5
Biosil, Degudent	64.8	28.5	5.3	0.5	0.5	0.4		
SLM Co212-f ASTM F75	65.2	28.3	5.48	0.754			0.164	

The SLM samples (Fig. 1) were fabricated directly from the virtual 3D models using SLM125 machine of the “SLM Solutions”, Germany. The machine is equipped with continuous Nd:YAG laser which worked with power 100 W and laser spot diameter 0.2 mm. The metal powder of Co-Cr alloy Co212-f ASTM F75 (Table 1) was melted in layers with 0.03 mm thickness unless the desired construction was obtained. During manufacturing process the laser at first scanned the outer contour of the layer of the first specimen’s part, next it hatched the area within the boundaries at an angle of 45° with a pitch of 0.13 mm. After that it passed to the same layer of the next specimen’s part, thus fabricating the whole layer. The laser path in manufacturing of tensile test specimen is shown on Fig.2. The SLM technological regime, recommended from the company producer was used. The tensile test specimens have a thickness of 2.2 mm for cast alloy and 2.07 mm for SLM one, as well as width of 6 mm.

Mechanical properties characterization

The Vickers hardness (HV) was measured on preliminary prepared cross-sections of the bridges along depth of all their elements with 100 gf loading. The Rockwell hardness was measured in different areas of the polished tensile test specimens and the average value is taken. The microstructure of the cast and SLM bridges was investigated by optical microscopy.

Uniaxial tensile tests were performed using an *FM*-1000 testing machine at room temperature. The strain was measured using a strain gauge. Typical samples used throughout the study are shown in Fig. 3 and Fig. 4. The 0.2% offset yield strength (0.2%YS) and elastic modulus were obtained from the stress–strain curve. The fracture surfaces were evaluated after tensile tests using optical microscope.

3. Results obtained

Hardness

The Vickers hardness distribution along depth of the cast bridge is given on Fig. 5. It is uneven with high fluctuations of the values within the range of 222 HV0.1. The hardness distribution along depth of the SLM sample is much more even, as it is shown on Fig. 6 with lower average deviation of the values – 184 HV0.1. The average Vickers hardness of the cast samples is lower than that of the SLM ones: 335 HV0.1 and 382 HV0.1 respectively (Fig.7-a). The average Vickers hardness values of dental bridges are in very good agreement with the average Rockwell hardness, measured on the tensile test specimens. The Rockwell hardness (Fig. 7-b) of the cast sample is 33 HRC (331 HV), which is lower than that of the SLM samples – 39 HRC (382 HV). The Rockwell hardness measurements confirm the lower hardness deviation of the SLM samples.

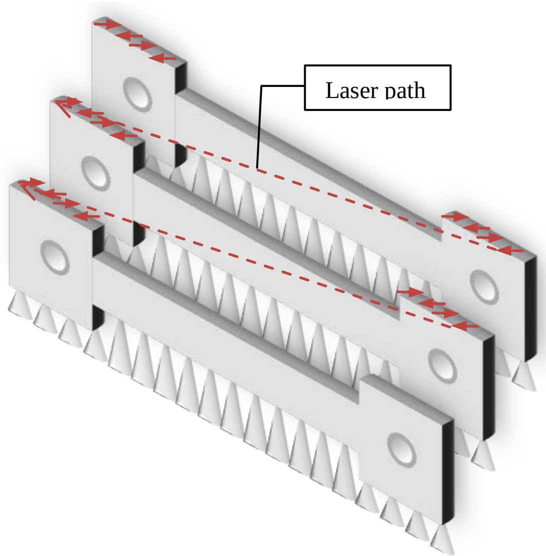

Fig. 2. *Scheme of tensile test specimens’ production by SLM*

Fig. 3. *Cast tensile test specimen with strain gauge.*

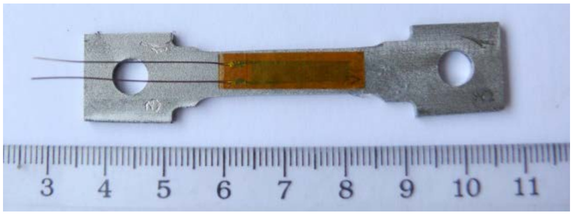

Fig. 4. *SLM tensile test specimen with strain gauge.*

Table 2.
Mean of mechanical properties of cast and SLM alloys.

Variable	Biosil F	SLM
Yield strength (MPa)	410	720
Elastic modulus (GPa)	209	213
Hardness HRC	33	39
Hardness HV 10	335	382

Tensile test

Figure 8 shows the typical nominal stress–strain curves of the SLM specimen and as-cast alloy. On the basis of such curves, the mechanical behavior can be recorded, including elastic deformation, and plastic deformation with the yielding process. Table 2 is a summary of their mechanical properties. The SLM specimens showed higher 0.2% yield strength compared to the as-cast alloy – 720 MPa and 410 MPa respectively. The yield strength of SLM specimens meets the standard ISO 22674:2006 for dental restorations (> 500 MPa).

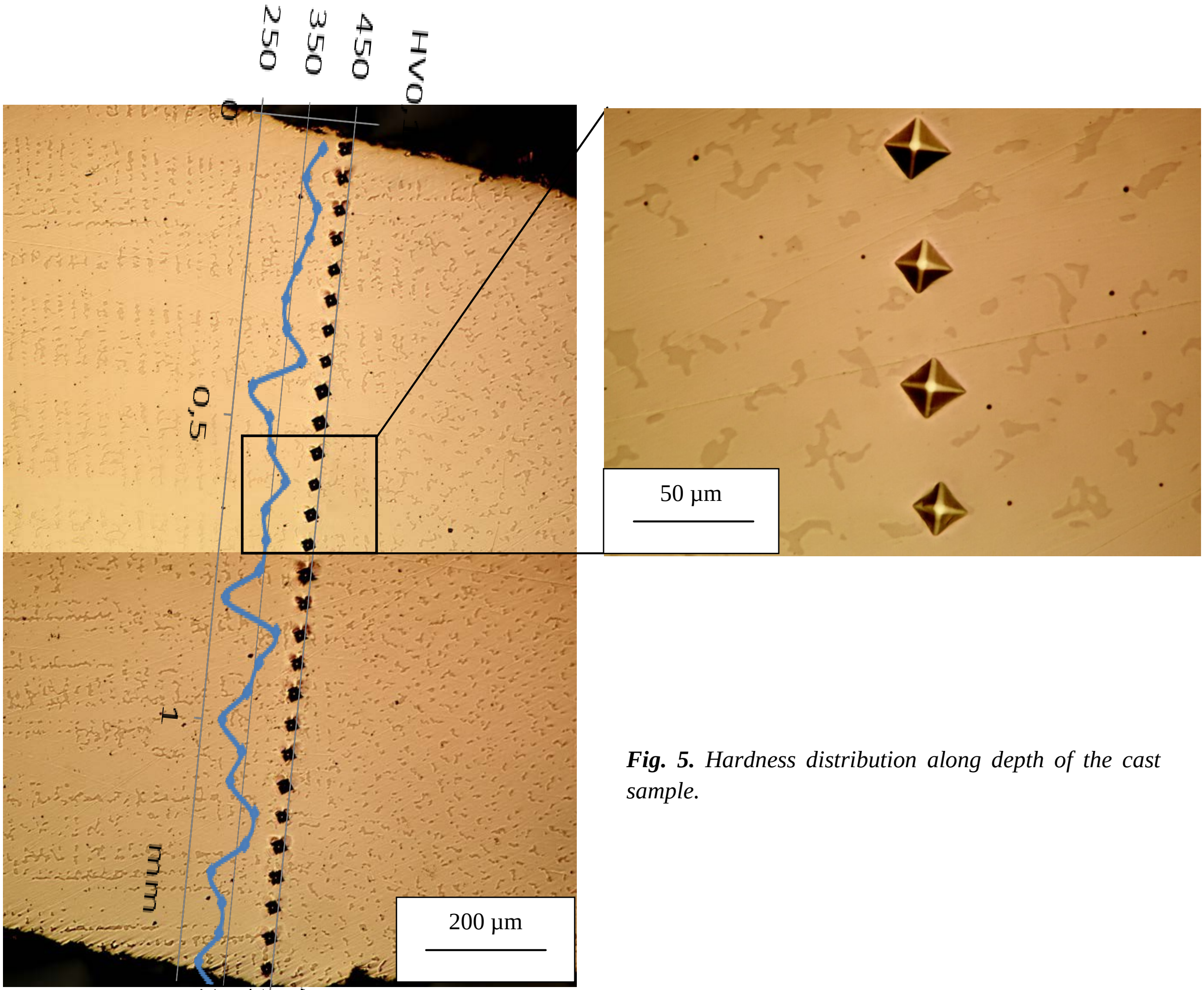

Fig. 5. Hardness distribution along depth of the cast sample.

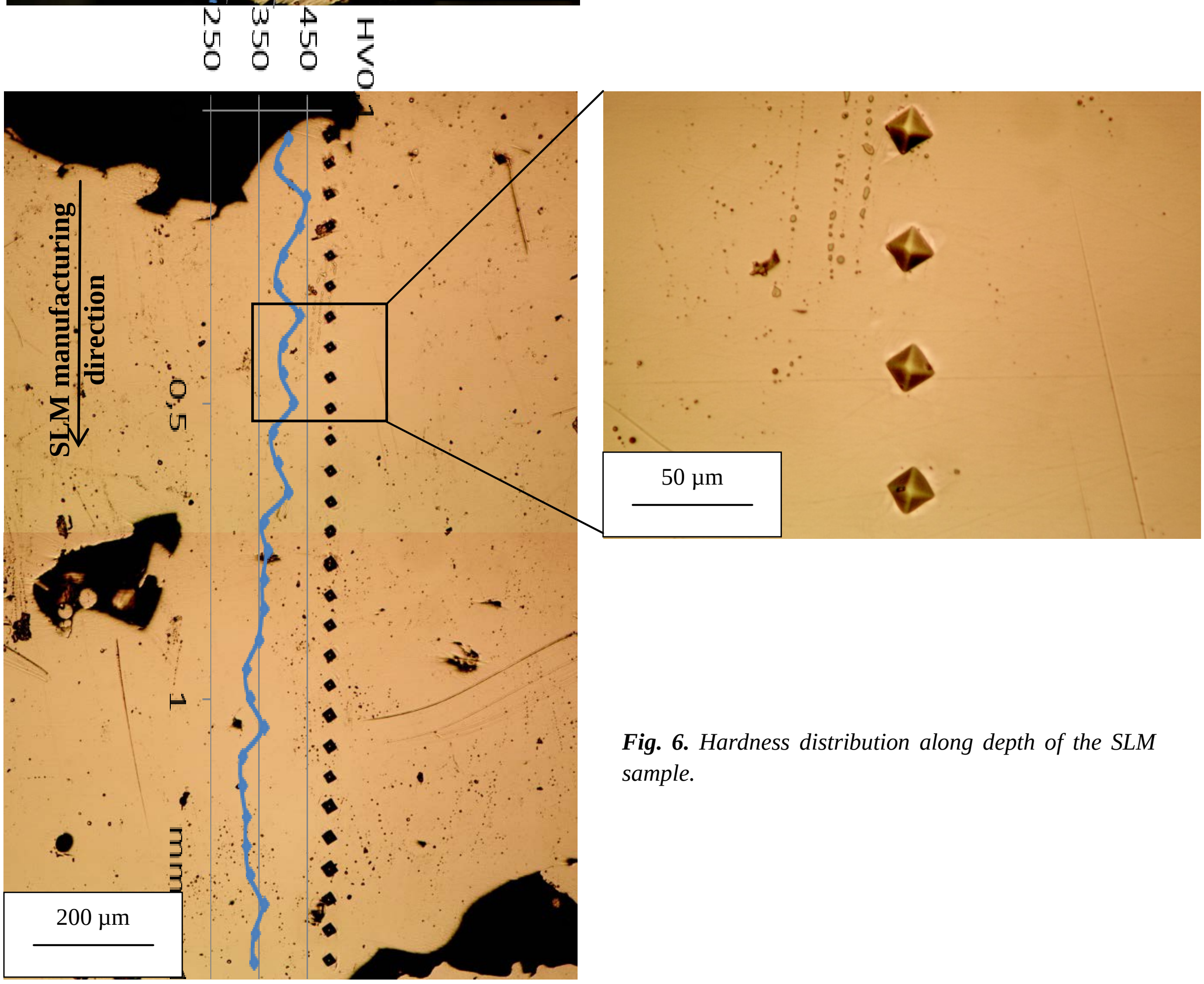

Fig. 6. Hardness distribution along depth of the SLM sample.

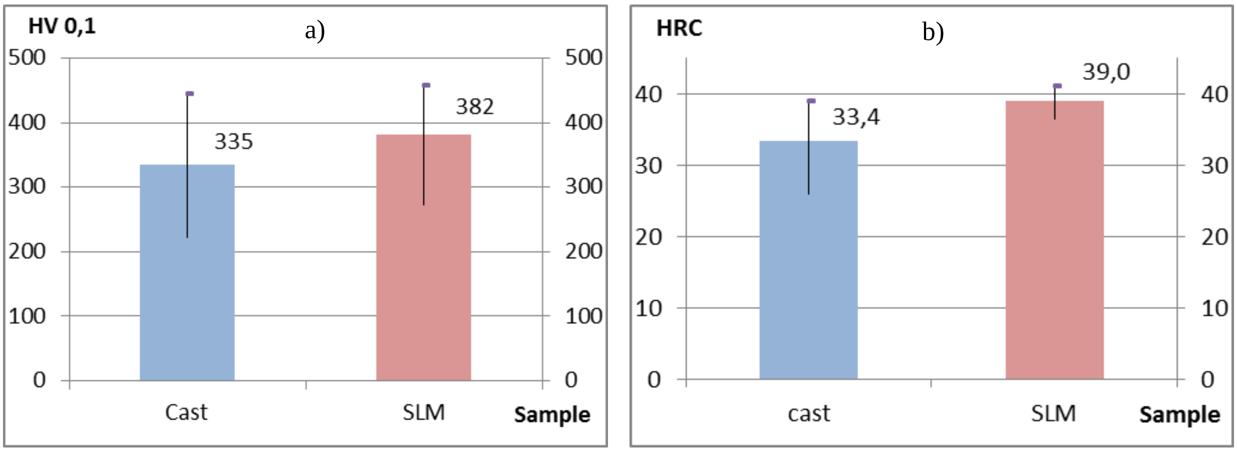

Fig. 7. Vickers hardness: a) and Rockwell hardness; b) of samples produced by casting and SLM.

4. Discussion

The properties of the Co-Cr-Mo alloys depend on the microstructure, its morphology and composition, γ-ε ratio, presence of carbides and intermetallic precipitations. Mechanical properties are affected by many factors, such as phase, grain orientation, grain boundary conditions, defects, etc.

The microstructure of the cast sample (Fig. 5) characterizes with dendrite morphology and carbides with round shape and small sizes, situated inside the grains. Our previous investigations [18] showed that the dendrites are composed of γ phase with lower strength, while the interdendritic regions consist of γ-phase/intermetallic eutectic with carbides, defining the higher hardness. The prevailing volume fraction of the lower strength γ phase is the main reason for the lower hardness of the cast samples, while the microstructural inhomogeneity is responsible for the higher fluctuations of the hardness values (Fig. 5).

The SLM process characterizes with high heating and cooling rates, leading to fine grained microstructure of the solidified layer. As the heat is lead away through the solid body, phase transformations run in the underneath layers heated above the transition temperatures, resulting in more homogeneous microstructure. Increasing the samples volume during the manufacturing process reduces the cooling rates which lead to slightly hardness decrease (Fig. 6).

The microstructure of the SLM sample (Fig. 6) possesses high corrosion resistance and could not be etched by immersion in any reagent, given in the references. Pores with different sizes, elongated along the direction of the layers melting, as well as unmelted powder can be observed in the whole volume. Our previous investigations [18] showed even distribution of the chemical elements in the dense areas of the SLM sample, proving the more homogeneous microstructure. In samples, produced by SLM, the higher hardness is attributed to the homogeneous microstructure with fine morphology [14] and the higher volume fraction of ε phase due to the incomplete γ-ε transformation [15], which is defined by the process peculiarities.

Mechanical properties of the SLM specimens and casting ones were investigated and compared. The SLM alloy possessed improved mechanical properties in comparison with that of the cast alloy. Yield strength and tensile strength of the SLM specimen was improved due to the unique microstructure. The higher tensile strength of the SLM specimens may be due to the finer grain size, cellular dendrite, and elongated precipitates. Also residual stress in the SLM specimen generates the residual strain [12] and affects on the tensile strength. Additionally, the presence of residual stresses during sintering is possible reason for the increased hardness in the SLM.

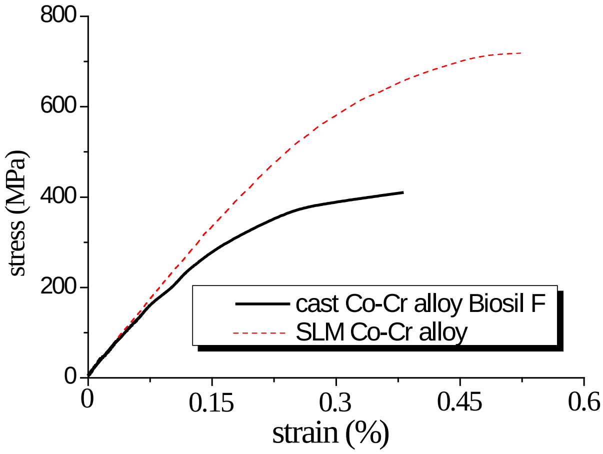

***Fig. 8.** Typical stress – strain curves for both specimen groups.*

Certain differences in the mechanical characteristics of the casting and SLM alloys can be related to its composition and the specifics of the manufacture processes. Though the composition of the alloys used for SLM and casting generally match however they differ by a small percentage of basic ingredients (Table 1). It has been shown that modification in the composition of an alloy can influence the mechanical properties by a large degree. SLM as a complex thermo-physical process can vary the final product depending on several factors, e.g., laser, scan and parameters of the environment. Changeable variables include laser power, layer thickness, scan speed and hatch spacing. These variables can be adjusted accordingly, optimizing some aspects that can have a negative effect on the mechanical properties of the material.

5. Conclusion

This study is focused on the microstructural and mechanical characterization of Co–Cr alloys fabricated using casting and selective laser melting. Within the limitations of this study, the following conclusions can be derived:

- Co–Cr dental alloys fabricated via casting or SLM techniques show significant differences in strength and hardness.
- The average Vickers hardness of the samples, produced by SLM, was higher than that of the cast samples – 382 HV and 335 HV respectively. The Rockwell measurements confirmed the higher hardness of the SLM samples – 39 HRC in comparison with that of the cast ones – 33 HRC.
- The SLM specimens showed higher yield strength compared to the as-cast alloy – 720 MPa and 410 MPa respectively.

Compared with the currently used cast alloy, the SLM alloy possessed improved mechanical properties. Co-Cr dental alloy fabricated with SLM is a promising alternative to conventional cast alloy for metal ceramic restorations.

6. References

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