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ORIGINAL ARTICLE

Fracture strength of porcelain fused to metal crowns made of cast, milled or laser-sintered cobalt-chromium

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Abstract

Aims. The aim was to compare the fracture strength of porcelain fused to metal crowns with copings fabricated in Co-Cr using different manufacturing techniques (casting, milling and laser-sintering) with crowns manufactured in a high-gold alloy. ***Methods.*** A total of 50 identical crowns were fabricated and sub-divided into five groups; cast Co-Cr, milled Co-Cr, two groups of laser-sintered Co-Cr and a control group cast in a high-gold alloy. After thermocycling (5000 cycles, 5–55°C) and pre-load (30–300 N, 10,000 cycles) the crowns were loaded until fracture. Load (N) and fracture mode were recorded. ***Results.*** There was a significant difference ($p < 0.05$) in fracture strength between the control group and one of the laser-sintered groups. The mean values (N) for the groups were as follows: cast Co-Cr, 1560 ± 274; milled Co-Cr, 1643 ± 153; laser-sintered Co-Cr 1, 1448 ± 168; laser-sintered Co-Cr 2, 1562 ± 72; control group, 1725 ± 220. ***Conclusion.*** There is no difference in strength between Co-Cr crowns produced using the different production technologies: casting, milling or laser-sintering. Metal ceramic crowns made with copings fabricated in a high-gold alloy present numerically higher fracture strength than crowns made with copings fabricated in Co-Cr alloys. The difference is confirmed when analyzing the fracture surfaces, but the difference in fracture strength value is limited and is only significant with regard to one of the two laser-sintered groups.

Key Words: *CAD/CAM dental, Co-Cr alloys, dental laser sintering, fracture strength, metal ceramic alloys*

Introduction

Although all-ceramic crowns are increasingly used in dentistry, porcelain fused to metal (PFM) is still considered to be a gold standard for crowns and fixed dental prostheses (FDPs) [1–7]. The main reason for using metal alloys for cores and frameworks for fixed dental prostheses is that they exhibit high strength and toughness and offer adequate support for the veneering porcelain [8–10]. Results from clinical studies show that the mean clinical survival rate of reconstructions made of PFM is 10–20 years or more [10–15] and that the reasons for replacing them are often other than mechanical failure of the reconstruction [1,3–5,10,13]. The majority of clinical studies relate to PFM based on high-gold alloys, gold being the dominant metal for PFM during the late 20th century. Today, however, high-gold alloys have been replaced to a great extent by other metal alloys for FDP frameworks, such as cobalt-chrome, due to the low cost and favorable mechanical properties of the latter [3,5,9,16–20].

Cobalt-chromium (Co-Cr) is relatively inexpensive compared to gold alloys and some all-ceramic materials and exhibits material properties considered suitable for dental reconstructions such as high strength, high modulus of elasticity (~ 250 GPa compared to 70–100 GPa for PFM high-gold alloys) and high corrosion resistance [16,17,21–24]. Co-Cr is also often described as a biocompatible material and has been used increasingly in recent years [2,5,6,16,19,23].

Co-Cr for dental purposes is manufactured using three different production technologies: the lost wax technique, CAD/CAM milling and direct metal laser sintering (DMLS) [25]. The different production methods are based on alloys that are specially developed for the specific production technique used.

Production methods for Co-Cr

Casting using the lost wax technique. A wax pattern is fabricated and invested in a phosphate-bonded investment material to create a mold for casting using the

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lost wax technique. Casting is usually carried out with induction heating in combination with either the centrifugal casting or vacuum-pressure casting technique. A modern alternative technology includes computer-aided design (CAD) and computer-aided milling (CAM) for the production of a burn-out pattern resin that is invested instead of the wax-pattern [20,23,26,27].

CAD/CAM technologies for Co-Cr. The manufacturing process is usually performed in three steps. First, data is captured either by scanning directly in the mouth (with an intra-oral scanner) or by scanning a cast (with a laboratory scanner) poured from an impression. Second, a digital production pre-shape is generated in the computer and finally a reconstruction is manufactured in the CAM-section [26–30]. During the last step, the virtual pre-shape serves as a pattern for milling a reconstruction from a solid Co-Cr blank or alternatively for directing a laser beam which melts Co-Cr granules to form a reconstruction. Using CAD/CAM technology in the production of dental reconstructions is both time-saving and cost-effective compared to conventional casting technology, which is often considered to be one of the reasons for using computer technology in dental production [20,23,26–28,30].

CAD/CAM technology: milling. CAD/CAM milling uses milling tools of different shapes and sizes to mill or grind reconstructions from solid blocks of materials [26–28,30]. Many different materials can be processed, e.g. titanium, various polymers, ceramic materials and Co-Cr [27,30]. One major advantage of using milling technology is that some disadvantages of casting, such as casting-induced flaws and porosities which can degrade the quality of the reconstructions, can be avoided as the blanks are manufactured under highly standardized industrial conditions [23,26,31].

CAD/CAM technology: direct metal laser sintering (DMLS). Laser sintering is a relatively new method compared to both conventional casting technology and CAD/CAM milling. During sintering, the Co-Cr or titanium powder granules are fused together by laser welding in layers using a high-energy carbon dioxide laser beam. As one layer is finished, the table where the finished part of the reconstruction is placed is lowered so that a new layer of green granules can be added over the sintered granules. Subsequently, the laser welds the new granules to the old ones, building up the reconstruction grain by grain until finished [23,30]. The technique allows for many prosthetic units to be produced at the same time and the powder granules not exposed to the laser beam can be re-used, both important advantages of the technology [23]. Furthermore, it was concluded in a study that the production technology is very precise and that the fit of a laser-sintered reconstruction is actually better than that of a reconstruction produced by CAD/CAM-milling [25]. Two other studies concluded that the marginal and internal fit of crowns produced using DMLS technology was comparable to that of crowns made using conventional casting techniques [7,20].

Different alloys that are adapted to each of the technologies are developed for the various production techniques mentioned above. Hence, it is difficult to draw general conclusions about the properties of Co-Cr based on only one of the alloys. Furthermore, the various production techniques result in reconstructions with different microstructures, such as grain size and surface morphology, which make comparison even more difficult [31].

One critical aspect of the clinical durability of PFM reconstructions is adaptation of the veneering porcelain to the metal substructure.

The material properties of the two materials that form the PFM laminate—the metal alloy and the porcelain—have to be compatible, especially in respect of the coefficient of thermal expansion (CTE) and chemical and mechanical behavior [2,6,8,10,19,21–23,31–36]. The intermediate oxide layer on the metal surface, the metal–porcelain interface, is also a crucial factor for clinical success which depends highly on the alloy composition [6,21,37].

Aim and null hypothesis

The aim of the present study was to investigate the fracture strength of PFM crowns based on different cobalt-chrome alloys in combination with each production method (cast, milled and laser-sintered) and to compare the results with those of a high-gold alloy-based control group. The null hypothesis was that there is no difference in fracture strength, regardless of which alloy or production method is used.

Materials and methods

Fifty identically-shaped crowns were fabricated and sub-divided into five groups of 10 depending on which material was used: cast Co-Cr, milled Co-Cr, two groups of laser-sintered Co-Cr of different brands and a control group made of a high-gold cast alloy.

Producing crown copings

A master die resembling a molar crown preparation was made of type IV stone (GC Fujirock® CP, GC Europe N.V., Leuven, Belgium) for the production of the 50 copings. A silicone impression (Dublisil 15, Dreve Dentamid GmbH, Unna, Germany) was taken of the master die to allow for fabrication of a die replica. The die replica was then scanned twice using

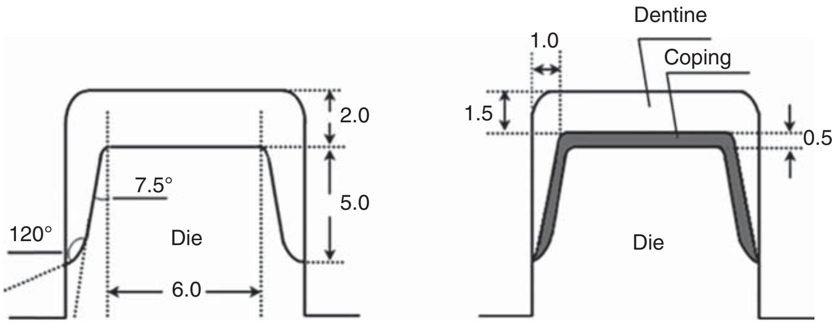

Figure 1. Design and dimensions of the standardized crowns.

two laboratory scanners, once for the manufacture of casting pre-shapes made from a burn out pattern resin for the cast groups (Everest Scan Pro, Kavo Dental GmbH, Biberach, Germany) and once for the milled and laser-sintered groups (3Shape D700, 3Shape A/S, Copenhagen, Denmark). The shape of the copings was designed with identical dimensions for all five groups (Figure 1).

Producing cast Co-Cr and the cast control group

The Everest Scan Pro CAD data was sent to a communicating milling machine (Everest Engine, Kavo Dental GmbH, Biberach, Germany) for the fabrication of 20 cast templates milled from a C-Cast disc (Everest C-Cast, Kavo Dental GmbH). The cast templates where then mounted five-by-five to the base of four sprue formers in total. The sprue formers were invested in a phosphate-bonded investment material (GC Stellavest, GC Europe, Leuven, Belgium) according to the manufacturer's instructions and two of them were used to cast 10 Co-Cr copings (Wirobond®280, BEGO Bremer Goldschlägerei Wilh. Herbst GmbH & Co. KG, Bremen, Germany) using an automatic vacuum and pressure casting machine (Nautilus® CC, BEGO, Bremen, Germany). Cooling procedure, deflasking and blasting with 250 μm aluminum oxide at a pressure of 3 bar and 20 mm distance between nozzle and specimen surface with an angle of 45° were all carried out according to the manufacturer's instructions.

The other two sprue formers were used to cast the high-gold alloy (Sjödings M2, K. A. Rasmussen AS, Hamar, Norway) using the same casting machine according to the manufacturer's instructions. Five of the copings were cast with 100% new alloy and the rest of the copings were cast with 50% surplus of the same high-gold alloy. The copings were sandblasted with 125 μm aluminum oxide at a pressure of 2 bar at a distance of 20 mm from a specimen surface with 45° angel.

Producing milled Co-Cr and the two groups of laser-sintered Co-Cr

The 3Shape CAD data was sent to a production center (M-tec Dental AB, Malmö, Sweden) where the copings were milled from Co-Cr alloy blocks (Gialloy CB, BK Giulini GmbH, Ludwigshafen, Germany). The same data was also sent to the two production centers where the laser sintering was to be performed using direct metal laser sintering technology (DMLS). At one of the centers (Biomain AB, Helsingborg, Sweden), copings were sintered from EOS CobaltChrome SP2 granules (Biomain AB) and subsequently sandblasted with 125 μm aluminum oxide at a pressure of 3 bar by the manufacturer before delivery. At the other center (BEGO Dental, Bremer Goldschlägerei Wilh. Herbst GmbH & Co. KG, Bremen, Germany), copings were sintered from Wirobond® C^+ cobalt chrome granules (BEGO Dental, Bremer Goldschlägerei Wilh. Herbst GmbH & Co.). The manufacturer sandblasted the copings before delivery with 250 μm aluminum oxide at a pressure of 3 bar (Table I).

Table I. Chemical composition of alloys as a percentage according to the manufacturer's instructions.

Alloys	*Co*	*Cr*	*Mo*	*W*	*Ga*	*Si*	*Mn*	*Fe*	*C*	
Wirobond® 280 (Cast)	60.2	25	4.8	6.2	2.9	*	*	—	—	
Gialloy CB (Milled)	61.7	27.50	—	8.71	—	1.57	0.26	0.14	0.04	
EOS CobaltChrome SP2 (DMLS 1)	62–66	24–26	5–7	4–6	—	0.8–1.5	*	*	—	
Wirobond®C^+ (DMLS 2)	63.9	24.7	5	5.4	—	1.0	—	—	—	
	Au	Pt	Ir	Pd	Ag	Cu	Zn		In	Sn
Sjödings M2 (Control)	84	7.90	0.10	5	0.90	0.30	0.10	0.20	1.0	0.50

Trace.

Metal surface preparation

All copings were to be veneered with the same porcelain, Noritake Super Porcelain EX-3 (Noritake Dental Supply Co., Limited, Aichi, Japan), and were cleaned with acetone (Acetonum, EurKval E, Gothenburg, Sweden) in an ultrasonic cleaner for 10 min before veneering according to the recommendations of the porcelain manufacturer. Oxidation was then performed, this time according to the metal manufacturer's instructions. The copings were then refinished and steam cleaned before applying the porcelain.

Porcelain build-up

The master die was used to hold and support the crown copings in a reproducible position during porcelain build-up. The shape and dimensions of the veneer build-up were determined by using a specially-made knife to shape the porcelain, as described in a previous study [38].

One layer of Paste Opaque Base Paste (PoBA, Noritake Dental Supply Co., Limited, Aichi, Japan) was applied to Co-Cr copings and fired in a calibrated porcelain furnace (Programat P500, Ivoclar Vivadent®, Schaan, Liechtenstein). Then one layer of Paste Opaque (Paste Opaque, Noritake Dental Supply Co., Limited) was applied to the copings and fired. In the control group, two layers of the Paste Opaque were applied to the copings according to the manufacturer's recommendations. Two dentine firings were carried out. In a final step, the crowns were autoglazed. All firing cycles were carried out according to the manufacturer's recommendations.

Support and cementation

Fifty resin dies were made for supporting the crowns during thermocycling, pre-cycling and load to fracture by casting Duralay® inlay pattern resin (DuraLay; Reliance Dental MFG Co., Worth, IL) in silicone impressions made of the master die according to a previous study [38]. All crowns were cemented on the Duralay dies using a self-curing luting composite cement (Multilink® Automix, Ivoclar Vivadent AG) under a standardized load of 15 N. Excess cement was removed and the crowns were stored in a moist environment at a temperature of 37°C to prevent desiccation of the cement until they were subjected to thermocycling and pre-loading.

Thermocycling

All crowns underwent 5000 thermocycles prior to the pre-loading procedure in an LTC 100 thermocycling device (LAM Technologies Electronic Equipment, Firenze, Italy) with two water baths at 5°C and 55°C. Each cycle lasted for 60 s: 20 s in each bath and 10 s to complete the transfer between the baths.

Pre-loading and load until fracture

All crowns underwent cyclic pre-loading for 10,000 load cycles with a load profile in the form of a sine wave at 1 Hz and min/max loads of 30 N and 300 N. The load was applied with a stainless-steel ball with 4.0 mm diameter placed on the occlusal surface of the crowns. All crowns were mounted at 10° inclination relative to the long-axis of the crown and submerged in distilled water during and in between the tests.

In a second step, all crowns were mounted in a testing jig at a 10° inclination relative to the long-axis of the crowns and finally loaded until fracture occurred using a universal testing machine (Instron 4465, Instron, Canton, MA). The load was again applied with a 4.0 mm diameter stainless steel ball placed on the occlusal surface of the crowns and the crosshead speed was 0.255 mm/min. A plastic foil with a thickness of 1.0 mm (Erkoflex, Erkodent®, Pfalzgrafenweiler, Germany) was placed between the occlusal surface and the stainless steel ball. The loads at fracture were registered (N) and any differences ($\alpha = 0.05$) between groups were calculated using one-way ANOVA followed by Tukey's post-hoc test.

Analysis

All fracture surfaces were analyzed using a light microscope (Wild M7A, Heerbrugg AG, Gais, Schweiz) in two magnifications (10 × and 20 ×). Subsequently two representative surfaces, one cast Co-Co and one control, were also analyzed using a scanning electron microscope (Quanta™ scanning electron microscope FEG-ESEM, FEI company, Eindhoven, Holland).

Results

The results are listed in Table II. The control group showed the highest fracture strength value among all the groups. In the Co-Cr groups, the milled group showed the highest mean value while the laser-sintered group (DMLS 1) showed the lowest. The differences between the groups, however, were not significant except for the control group and one of the laser-sintered groups (DMLS 1, $p < 0.05$).

Analysis of the fracture surface

All crowns showed cohesive or mixed fractures and consequently no crown showed any adhesive fracture (Table III, Figures 3A and B). Visual inspection of the fractures, however, revealed that the exposed metal surfaces of the control group showed more evident ceramic residues compared to the Co-Cr metal surfaces where the metal was free from such residues.

Inversely, SEM surface analysis showed metal (Co and Cr) in the surface of the cast Co-Cr specimen, but no metal trace present in the surface of the control,

Table II. Load at fracture (N) for all groups.

Specimen number	Cast, Wirobond®280	Milled, Gialloy CB	DMLS 1 EOS CobaltChrome SP2	DMLS 2 Wirobond®C$^+$	Control Sjödings M2
1	1457	1421	1419	1513	2163
2	1997	1788	1387	1434	1972
3	1441	1785	1313	1546	1805
4	1997	1771	1213	1617	1780
5	1708	1794	1491	1509	1781
6	1440	1544	1544	1540	1654
7	1575	1612	1802	1693	1521
8	1490	1765	1313	1620	1497
9	1125	1484	1585	1555	1544
10	1365	1469	1408	1596	1533
Mean ± SD	1560 ± 274	1643 ± 153	1448 ± 168	1562 ± 72	1725 ± 220

only chemical components that could be related to the veneer material (Figures 4A and B).

Discussion

The present study investigated the fracture strength of PFM-crowns based on Co-Cr produced using different manufacturing techniques: casting, CAD/CAM milling and two DMLS techniques. The results only showed a difference in fracture strength between the control group and one of the DMLS-groups ($p < 0.05$). Despite the fact that there were no significant differences between the other groups, there are still numerical differences that deserve to be discussed, especially concerning the different chemical composition and different surface structure of the materials under study. Furthermore, the producers of the alloys/copings recommend different surface treatments prior to porcelain firing which might be an influencing factor on the results in the study.

A high-gold alloy was chosen for the control as this type of alloy often is considered to be the gold-standard for PFM-based reconstructions [31]. Furthermore, the Sjödings M2 has been commercially available for many years without change in composition and is compatible with the porcelain used in the study, a porcelain that has been developed to be compatible with both Co-Cr and high-gold alloys. The coefficient of thermal expansion (CTE) of the porcelain is $12.4 \times 10^{-6}/K^{-1}$ and the CTE for the different alloys used is within range ($14.0 \times 10^{-6}/K^{-1}$ for Wirobond®280, $14.1 \times 10^{-6}/K^{-1}$ for Gialloy CB, $13.9–14.3 \times 10^{-6}/K^{-1}$ for EOS CobaltChrome SP2, $14.1 \times 10^{-6}/K^{-1}$ for Wirobond® C$^+$, $14.3 \times 10^{-6}/K^{-1}$ for Sjödings M2) of what is recommended for the porcelain. The porcelain/alloy combinations used are recommended by both the porcelain manufacturer and the alloy manufacturers, an important factor with respect to the different materials used. The chemical composition of the different alloys can be one factor that influences the bond strength in the PFM-laminate. The different components in the alloy are added to impart specific properties to the material. Among those components, there are some that are important for the strength of the porcelain, such as oxide formers that build the oxide layer during initial oxidation prior to porcelain firing [8]. The oxidation process then continues during the subsequent firing cycles. The oxide layer allows for chemical bonds when the porcelain at the interface partly dissolves and saturates the metal oxide. The metal oxide, on the other hand, is saturated by metal, forming a continuous structure from the metal through the oxide layer and to the porcelain, chemically bonding the porcelain to the metal [2]. If, however, the oxide layer gets too thick, it will weaken the reconstruction because functional loads are transferred to the interface and particularly to the very brittle and weak oxide layer [21,37]. Hence, it was concluded in a study that bond strength of dental porcelains to different CoCr-alloys is dependent on the alloy composition, which might be one factor responsible for the differences in results between the groups [6].

The mechanical response of the functionally loaded PFM-laminate is furthermore dependent on the mechanical properties of the supporting metal core. Those mechanical properties are in turn dependent on material composition and production history, both factors that vary between the different groups. The material composition as determined by the various bonds that are formed between different atoms and molecules in the material and the production history, such as cooling rates, pressure and time, leads to a different microstructure within the material, all of which results in an altered mechanical response when loaded. In addition to this, the porcelain support may vary, leading to different strengths of the various materials under study. In the present study, however, the cores were shaped identically to exclude this as a factor that might influence the result.

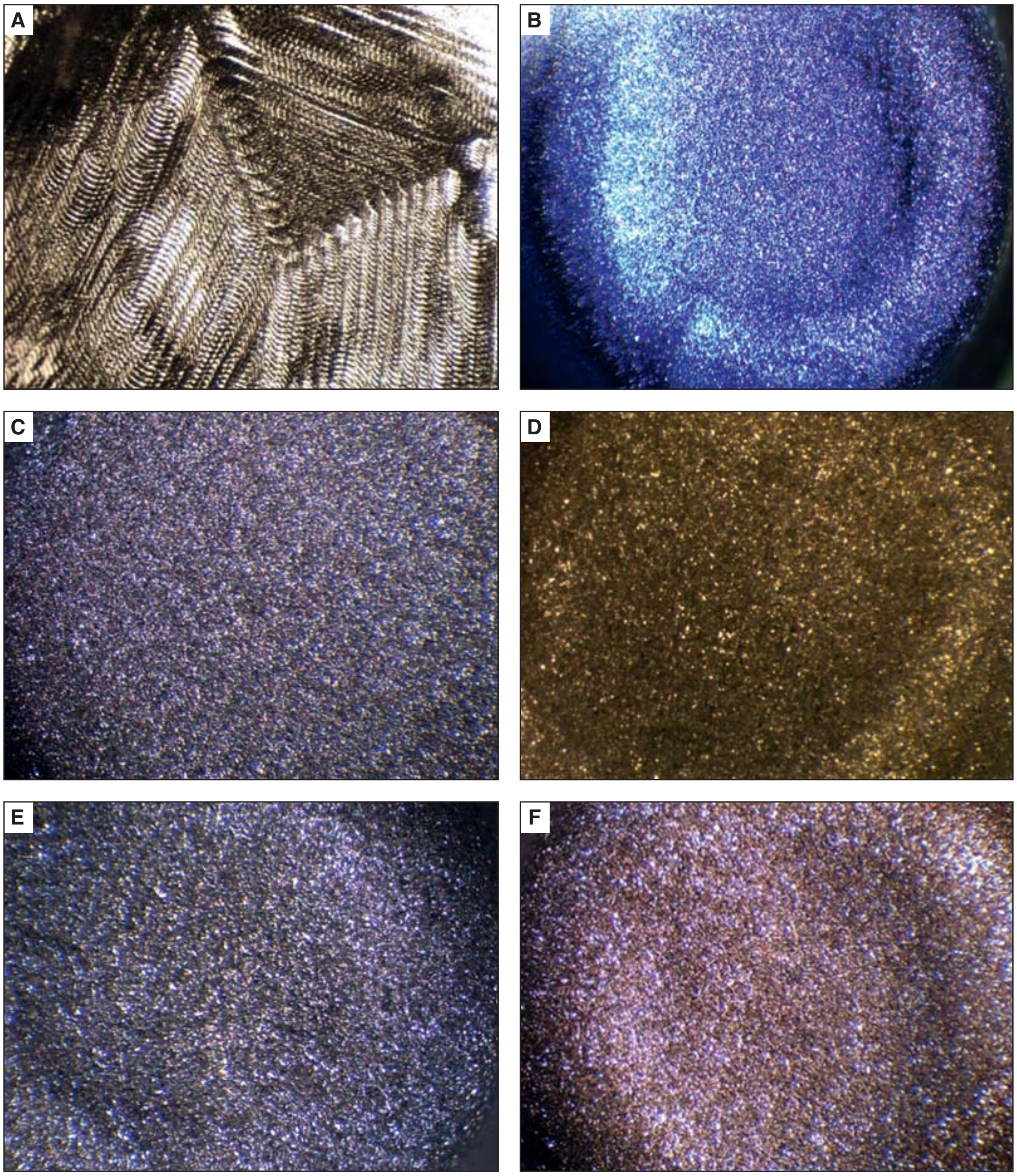

Figure 2. The metal surface of the copings prior to porcelain firing. (A) Milled Co-Cr before blasting. (B) Milled Co-Cr after blasting. (C) Laser-sintered Co-Cr (Biomain) after blasting. (D) Laser-sintered Co-Cr (BEGO) after blasting. (E) Cast Co-Cr after blasting. (F) Cast hi-gold alloy after blasting.

The surface treatment used for the various groups differed because the manufacturer's recommendations varied between the different materials. The grain size of the particles used for airborne-particle abrasion and the air-pressure used can be factors influencing the result. The suggested rationale behind the use of airborne-particle abrasion is that it might increase the metal/surface energy, improving the porcelain's wettability and, consequently, the bond strength due to micromechanical bonding [5,32,33,35,39–41]. If, on the other hand, the surface roughness is too great and irregular, this might result in small gaps in the surface structure favoring bubbles during porcelain build-up, resulting in flaws in the interface and thus decreasing strength [39]. In the present study, the same pressure of 3 bar was used for all the Co-Cr specimens, but the grain size varied according to the alloy manufacturers' suggestions. For the DMLS 1 group and the milled group, 125 μm Al_2O_3 was used, while for two other Co-Cr groups (DMLS 2 and cast Co-Cr), 250 μm Al_2O_3 was used (Figure 2).

There was no significant difference between results in the Co-Cr group that could be explained by differences in the surface treatment of the metal prior to porcelain application. When comparing the results with the control, the Sjödings M2 group, the pressure used was lower (2 bar), but grain size (125 μm Al_2O_3) was identical to two of the Co-Cr groups. There is a substantial difference between the hardness of gold and Co-Cr which is the main reason for using a different air pressure due to the fact that a higher pressure is needed for the harder Co-Cr. When

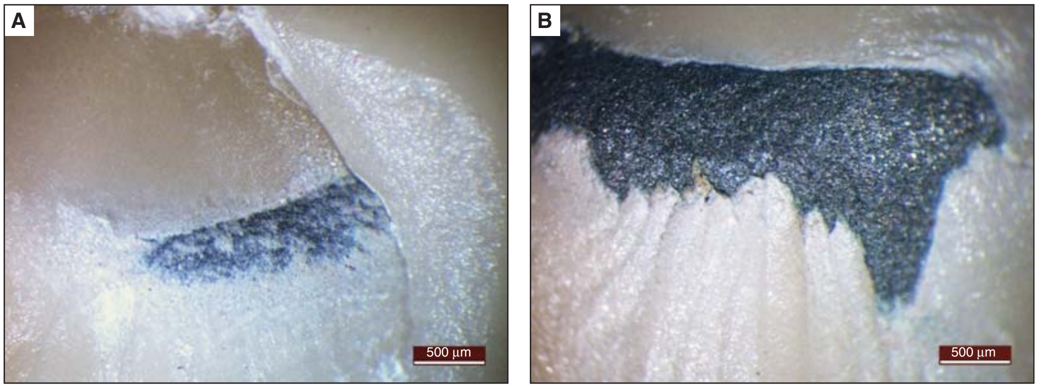

Figure 3. Representative of typical fracture modes after fracture strength test (× 20). Mixed fracture representative of the control group with residues of porcelain present on the alloy surface (A) and mixed fracture representative of the Co-Cr groups (B).

comparing the results, there is one significant difference between two of the groups: the DMLS 1 and the control. If the DMLS 1 group had been treated with 250 µm Al_2O_3 air-particle abrasion instead of 125 µm Al_2O_3, the results of this group might have been higher and within the range of the results in the DMLS 2 group, where no significant difference could be found between this group and the Sjödings M2 group. This supposition, however, could not be proven in this study, since there were more factors than air-borne abrasion that differed between the groups. The angle and distance of the nozzle during surface treatment are other factors that might influence the surface structure [39]. However, this was standardized in the study.

Occurring fracture types are either adhesive, cohesive or mixed. Fractures are defined as adhesive if occurring between the alloy and the oxide layer or between the porcelain and the oxide layer; the latter if the opaque layer is considered to be part of the porcelain. Those fractures may be due to a weaker bond between the different layers, for example in the interface between the porcelain and the alloy, by comparison with the bonds within the materials. The cohesive fractures occurred within a material: the porcelain, the metal or the oxide layer. In this case, the inherent bonds are weaker than the bonds in the interfaces. The third type of fracture is a combination of cohesive and adhesive fracture known as mixed fracture [8,16,23,32].

Most of the fractures in the present study were of the mixed type; no crowns showed adhesive fractures irrespective of group. Residues of porcelain were observed on the alloy surface in the nine cohesively-fractured crowns in the control group (Figure 3A). When analyzing the chemical composition of two representative metal surfaces, this observation was confirmed. Since high concentrations of Co and Cr

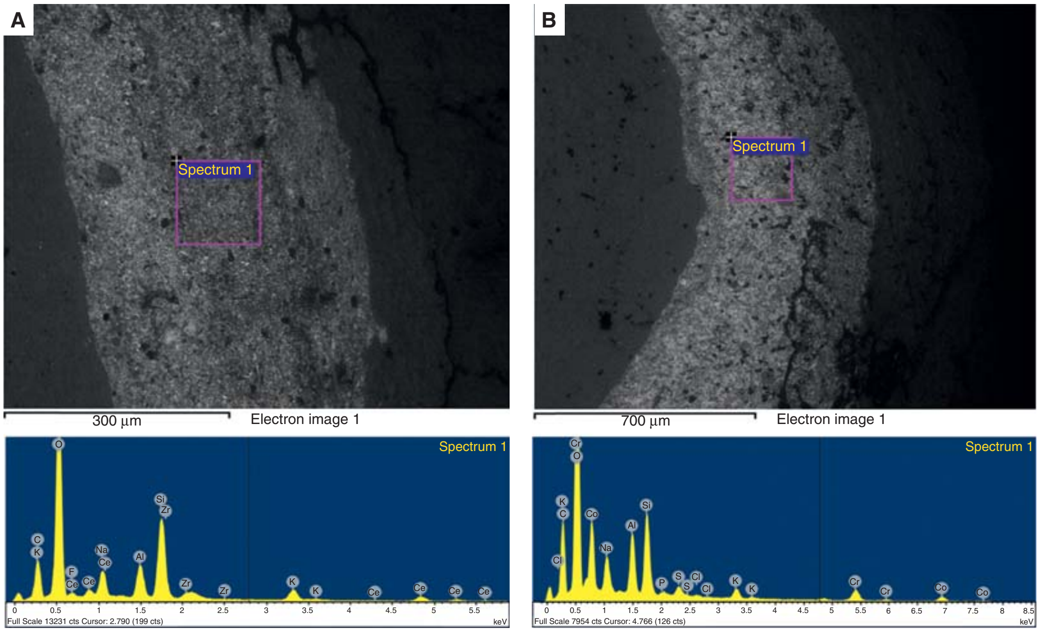

Figure 4. Scanning electron microscope (SEM) analysis of the fracture surface after fracture showed no exposed Au in the surface of the control (A) but traces of both Co and Cr in the surface of the cast Co-Cr specimens (B). This might be explained by ceramic residues covering the gold surface while the fracture surface of the Co-Cr specimens show more free metal exposed.

Table III. Fracture mode.

Group	Adhesive fracture	Cohesive fracture	Mixed fracture
Cast (Wirobond®280)	—	3	7
Milled (Gialloy CB)	—	2	8
DMLS 1 (EOS CobaltChrome SP2)	—	1	9
DMLS 2 (Wirobond®C$^+$)	—	2	8
Control (Sjödings M2)	—	1	9

were found in the analyzed surface of the Co-Cr sample, but only traces of veneer material and the oxide layer could be seen in the control surface, the SEM analysis confirms that the bond between the metal and porcelain are likely to be stronger in the control group compared with the investigated Co-Cr materials. The clinical importance of this finding is, however, not clear, since the differences in fracture load were limited.

There are few studies on bond strength between porcelain and Co-Cr as a function of the method of manufacture. Unlike most studies, the subject of the present study was to investigate the fracture strength between porcelain and Co-Cr due to the different production techniques. In a comparative study, Akova et al. [23] compared two different production techniques, laser sintering and casting technology, to examine the bond strength between laser-sintered Co-Cr and cast Co-Cr and Ni-Cr to porcelain. The authors concluded that there was no statistically significant difference between the groups and that the laser sintering technique for the Co-Cr alloy appeared promising for dental applications. The cast groups showed higher bond strength values than the laser-sintered group. The specimens in the study had a cylindrical shape which differs from the specimens in the present study. The authors performed neither thermo-cycling nor mechanical pre-loading of the specimens [23].

Five of the 10 copings in the control group were cast with 50% surplus of high-gold alloy. There was no significant difference in the fracture strength values of the crowns which were cast with a surplus of gold and those that were cast with 100% new gold alloy.

According to the porcelain manufacturer's instructions, two layers of Paste Opaque were applied to the crowns in the control group while one layer of base paste (POBA) and one layer of Paste Opaque were applied to the Co-Cr groups. It is difficult to determine how thick or thin an opaque layer or a POBA was applied to the copings. This is a factor that might have influenced the scatter of the results.

Aspects of the methods used

Duplicating the clinical environment in an *in-vitro* test set-up is challenging and could be questioned for several reasons. In the clinical situation, fatigue is important since dental restorations are subjected to alternating forces during mastication where water plays an important role in the development of slow crack growth [12,38]. The way of loading the specimens during pre-cycling and load until fracture is another factor that has been debated, since it is considered important to produce fractures of the same mode as can be seen in the clinical situation. Tests on standardized crowns can be used to obtain helpful information for comparing different materials, as the shape of the reconstruction can influence fracture behavior [29,42].

Previous studies have used different die materials such as brass, Duralay® or extracted teeth to support the crowns to be tested [29,31,38,42–44]. If the standardized crown is cemented on a die that has a high modulus of elasticity, such as steel or brass, the fracture strength will be unrealistically high compared to a crown that is supported by a die with mechanical properties close to the natural tooth [38,43]. In the present study, the crowns were cemented onto abutments made of inlay pattern resin, Duralay®, because this material has mechanical properties close to the natural tooth [38,42]. However, as the test set-up used was the same for all the groups, it is reasonable to make a comparison between the different materials.

Conclusion

The following conclusions can be drawn within the limitations of the present study:

- There is no difference in strength between Co-Cr crowns produced using the various production technologies investigated in the study: casting, milling or laser sintering.
- Metal ceramic crowns made with copings fabricated in a high-precious gold alloy present numerically higher fracture strength than crowns made with copings fabricated in Co-Cr alloys. The difference is confirmed when analyzing the fracture surfaces, but the difference in fracture strength value is limited and is only significant in respect of one of two laser-sintered groups. Hence, the clinical importance of the difference cannot be verified and needs to be clarified by further studies.
- The null hypothesis was rejected.

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The authors report no conflicts of interest. The authors alone are responsible for the content and writing of the paper.

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