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Analytical and Electrochemical Evaluation of the *in vitro* corrosion behavior of Nickel-chrome and Cobalt-chrome casting alloys for metal-ceramic restorations.

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Abstract - *In this study we examined the hypothesis based on relevant literature survey that the in vitro corrosion behavior of a Cobalt-chrome dental casting alloy for metal-ceramic restorations is better than that of a Nickel-chrome dental casting alloy. The corrosion released metal ions were analysed by Inductively Coupled Plasma-Atomic Emission Spectroscopy. Moreover, the specimens were electrochemically tested by linear polarization. The statistical analysis of the results showed statistically significant differences in corrosion rates of Nickel-chrome alloy and Cobalt-chrome alloy calculated by analytical and electrochemical measurements. The hypothesis was confirmed and the results showed that the corrosion rates of the Cobalt-chrome alloy were lower than that of the Nickel-chrome alloy*

KEY WORDS: Corrosion, Nickel-chrome alloy, Cobalt-chrome alloy, metal-ceramic restorations

INTRODUCTION

The number and types of alloys for porcelain fused to metal (PFM) restorations has increased dramatically over the last 25 years. This fact made selection of an alloy for a given clinical situation difficult. The proper selection and manipulation of alloys is imperative if dental prostheses are to perform well over a long period[1]. There are 3 prime factors that influence the clinician's decision, when selecting a dental casting alloy for a specific clinical situation: 1) cost, 2) physical properties of the alloys, and 3) biocompatibility[2]. Unfortunately, cost factors have become more influential over the last 2 decades as a result of the rapidly fluctuating international cost of gold and noble metals, in general. However, the rationale for alloy selection should not exclusively be based on cost without consideration of the biological safety of the dental alloy. It is important to note that the oral environment is considered to be a harsh regime that promotes the corrosion of dental alloys. The oral cavity is moist and continually subjected to fluctuations in temperature. The food and liquids ingested have wide ranges of pH. The acids that are liberated during microbial fermentation of carbohydrates considerably reduce pH value. Furthermore, oral micro-organisms are organized into complex biofilms which adhere tenaciously to the metallic restoration providing a localized condition that is extremely conducive to an accelerated reaction between the oral media and the dental alloy.

Dental casting alloys are widely used in the applications that place them in contact with oral tissues for many years. The corrosion of dental alloys releases free aquated or complexed metal ions which may then come into contact and react with cells and tissues in the immediate environment, or be distributed throughout the human body. The issue of metal release from biomaterials implanted or inserted in various cavities of the human body has attracted the interest of many investigators because debris or degradation products can elicit an adverse immune response or induce pathologic processes[3,4].

Ni-Cr and Co-Cr casting alloys are part of the group of the predominantly base metal alloys[5] which now account for the majority of the prosthetic restorations used clinically. There is no doubt that these alloys are prone to various types of corrosion depending on the alloy composition and the oral environment. The *in vitro* corrosion behavior of Ni-Cr alloys for metal-ceramic restorations has been well investigated by researchers[6-12]. Nonetheless, the dental literature lacks a consensus on the corrosion resistance of Co-Cr alloys[13,14,15].

Any natural or artificial surface exposed in the oral cavity will be covered quickly by a noncellular salivary biofilm rich in acidic proteins and lipids, known as acquired pellicle. Human saliva can significantly interact with the teeth and the restoration biomaterials. The metal ions released from the metal-ceramic restorations are aquated as follows:

$$M^{n+} + 6.\ H_2O \rightarrow [M^{n+}(H_2O)_6]^{n+}$$

where M=metal and n=2,3,4, the charge of metal

The aquation process takes place by coordination of the water molecules which are in saliva almost 80% form normally an octahedral geometry (Figure 1).

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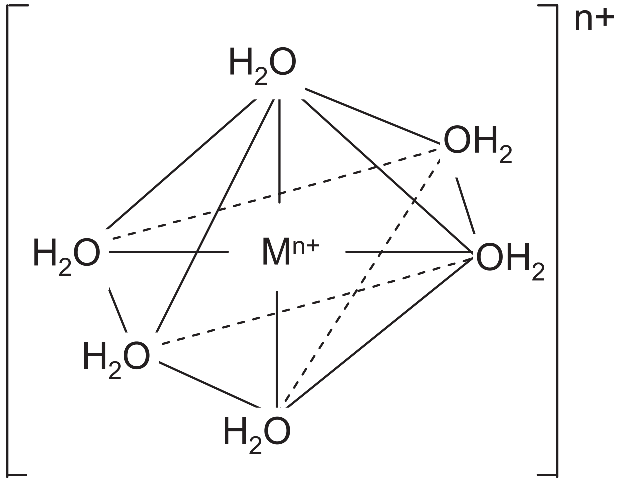

Figure 1. *Geometry of the aquation process.*

Table 1. *Chemical compositions of the test alloys according to the data provided by the manufacturer.*

Test alloy	*Element (wt%)*									
	Ni	*Cr*	*Co*	*Mo*	*W*	*Nb*	*Si*	*Fe*	*Ce*	*C*
Ni-Cr	65	22.5	-	9.5	-	<1	1	0.5	0.5	0.02
Co-Cr	-	26	61	6	5	-	<1	0.5	0.5	0.02

The aquation is as fast as a diffusion control reaction and is followed by complexation with the most electronegative ligand in saliva. The ligand may be an inorganic compound or ion, such as, phosphate, chloride or fluoride. The ligand could also be an organic like protein, amino acids, glucose, urea, uric acid, etc. The complex could be formed by a mixture of inorganic and organic ligands. For example, if M^{n+} is Cr^{3+} the reaction of complex is as follows:

$[C\,(H_2O)_6]^{3+} + L \rightarrow [C\,(H_2O)_5L]^{3+} + H_2O$

where L=ligand

AIMS

The aim of this study was to evaluate analytically and electrochemically the corrosion resistance of Ni-Cr and Co-Cr alloys for metal-ceramic restorations. The hypothesis tested in this study was that the *in vitro* corrosion behaviour of a Co-Cr casting alloy for metal-ceramic restorations is better than that of a Ni-Cr casting alloy for metal-ceramic restoration.

MATERIALS AND METHODS

Two different representative, commercially available dental casting alloys, designated as Wiron99/Bego (Ni-Cr alloy) and Wirobond C/Bego (Co-Cr alloy), were used as test specimens. 20 rectangular wax patterns (Kronenwachs/Bego) were prepared to produce cast specimens. Each wax pattern was sprued and invested individually in a graphite-free, phosphate-bonded investment material (Bellavest T/Bego). Then the wax patterns were fired according to manufacture's instructions. The preheating temperatures for W99 and WBC are 900-1000 ^{0}C. Casting with the above mentioned alloys was performed by using a vacuum pressure casting machine with induction melting device (Nautilus T/Bego). According to the recommendations from the alloy manufacturers the casting temperature of W99 and WBC are approx. 1420 ^{0}C and 1470 ^{0}C respectively. Then, the metallic specimens with dimensions of 17 x 10 x 1 mm were carefully deflasked, blasted with aluminum oxide (Korox 250/Bego) with grain size of 250 μm and working pressure from 4-6 bar, mechanically wet ground to P1200 silicon carbide paper (Nikken, Nihonkenshi Co., Ltd. Japan) removing at least 0.1 mm from all sides and finally polished with diamond paste (0.25 μm). After polishing, the specimens were cleaned ultrasonically in acetone for 5 minutes and rinsed with distilled water before corrosion tests.

Electrochemical Corrosion Tests

Three specimens of each alloy were immersed in an artificial saliva solution of 0.1 M NaCl and 0.1 M lactic acid with a pH of 2.3 [16,17]. All test specimens were stored in sealed containers. Open circuit potential (OCP) of each alloy was monitored as a function of time (total monitoring time: 1400 h). Another set of three specimens of each alloy was subject to linear polarization (LP. Triplicate corrosion tests for each alloy were performed using appropriate electrochemical setup (EG&G Princeton Applied Research Potentiostat/Galvanostat Model 362, Princeton, NJ, USA). The working electrode was each test alloy, the reference electrode was a standard calomel electrode (SCE), and the auxiliary electrode was platinum. Each specimen was polarized from -20 mV to 20 mV, at a scanning rate of 0.3 mV / s.

The polarization tests were repeated at least 3 times for each specimen until all repeated runs exhibited similar polarization curves. The corrosion rate was estimated with the following equation:

$$CR_{LP} = 0.82 \times I_{CORR} \times EW$$

where CR_{LP} (µg/d.cm²) stands for corrosion rate, EW is an equivalent weight of the corroding species in grams (g) and I_{CORR} is a corrosion curent density (µA/cm²). The corrosion current density (I_{CORR}) was obtained graphically by finding the intersecting point of the cathodic Tafel slope and the anodic Tafel slope. The means and standard deviations of the corrosion rates were obtained from 3 runs per each alloy. Student's t-test was performed at a 95% overall confidence level to identify statistically significant differences in corrosion rates of the alloys.

Inductively Coupled Plasma-Atomic Emission Spectroscopy (ICP-AES)

Three specimens of each alloy were immersed for 28 days in an artificial saliva solution of 0.1 M NaCl and 0.1 M lactic acid with a pH of 2.3. All test specimens were stored in sealed containers. At the end of the immersion period, 5 mL of eluent was removed from each artificial saliva solution by using a syringe with a plastic tip. The released metal ions were analysed by ICP-AES (Jobin-Yvon 138 ultrace, Lyon, France). Calibration of the apparatus was achieved with standard solutions, and computer-controlled data collection was used to assign peaks to specific elements. Corrosion rates of main components (Ni, Co, Cr, Mo) and the corrosion rate of each alloy were calculated. The means and standard deviations of the corrosion rates were obtained from 3 runs for each main component and alloy. Student's t-test was performed at a 95% overall confidence level to identify statistically significant differences in corrosion rates of the alloys.

RESULTS

Electrochemical corrosion tests

After 17 days of immersion stabilized corrosion potentials (E_{corr}) were observed: 220 mV for Ni-Cr (Fig.2) and 280 mV for Co-Cr (Fig.3) referred to Ag/AgCl electrode.

A mean corrosion rate of 0,139 µg/d.cm² for Co-Cr alloy and 0,224 µg/d.cm² for Ni-Cr alloy were estimated by linear polarization measurements, respectively. The statistical analysis of the results identified statistically significant differences in corrosion rates of Ni-Cr alloy and Co-Cr alloy ($p<0.05$).

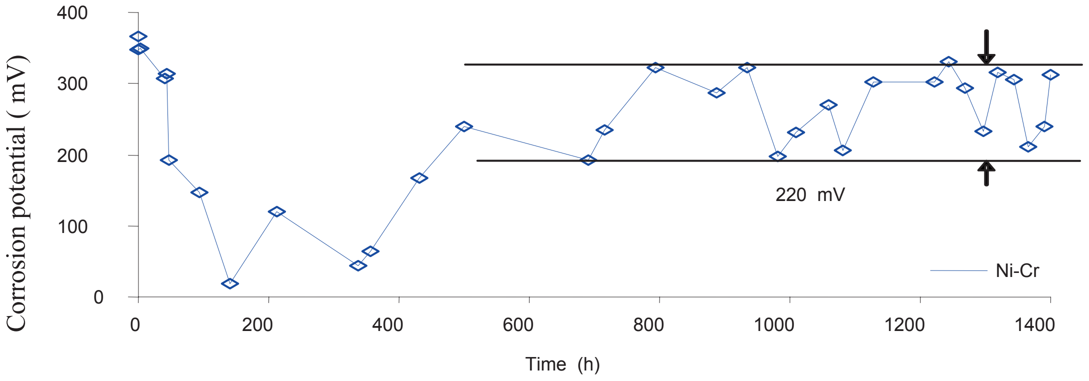

Figure 2. *Representative OCP vs time curve for Ni-Cr alloy (W99)*

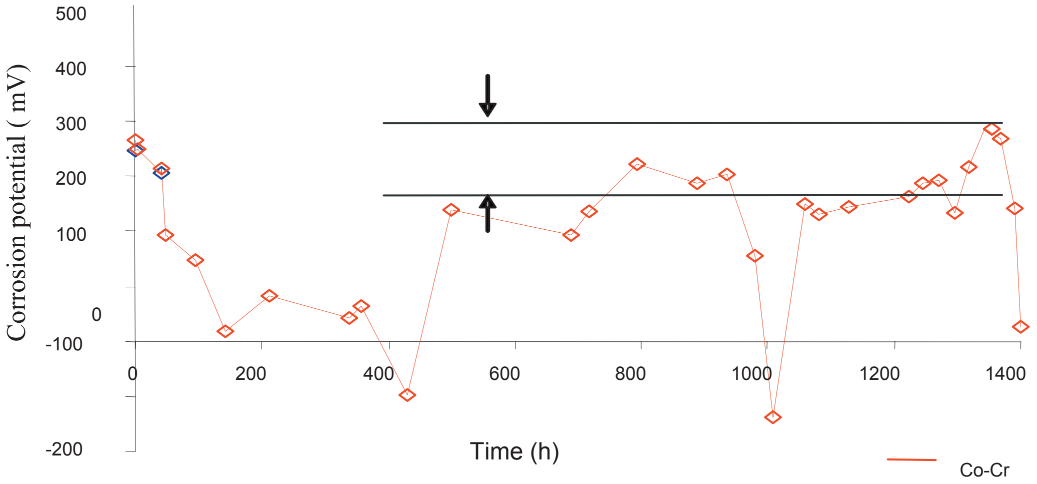

Figure 3. *Representative OCP vs time curve for Co-Cr alloy (WBC)*

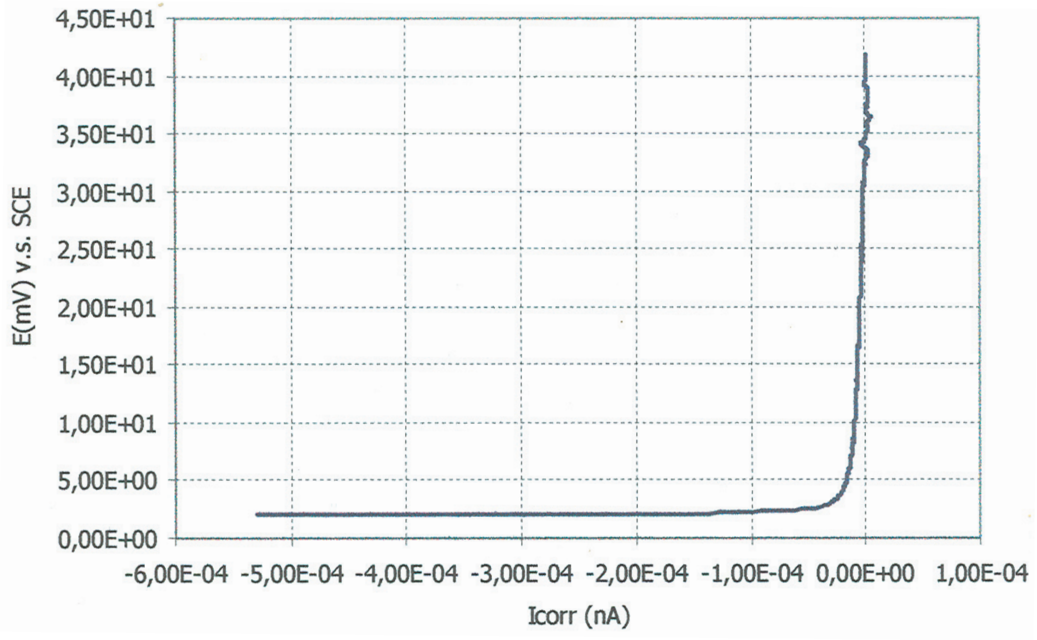

Figure 4. *Representative LP curve of Co-Cr alloy.The Ni-Cr alloy produced a similar LP curve.*

Table 2. *Element release in ppb (Mean ± SD) from the casting alloys W99 and WBC in the artificial saliva solution (0.1 M NaCl and 0.1 M lactic acid, pH = 2.3)*

Alloy	*Immersion period*	*Ni*	*Co*	*Cr*	*Mo*	*Total ion release (ppb)*
Ni-Cr (W99)	28 d (days)	8,5 ± 0,7	-	1,0 ± 0.3	2,0 ± 0,2	11,5± 0,2
Co-Cr (WBC)	28 d (days)	-	4,2 ± 0,3	1,0 ± 0.2	1,5 ± 0,5	6,7± 0,2

Table 3. *Mean corrosion rates (μg/d.cm²) of Ni, Co, Cr, Mo and each alloy.*

	Mean corrosion rate ± Standard Deviation (μg/day.cm²)				
Alloy	*Ni*	*Co*	*Cr*	*Mo*	
Ni-Cr (W99)	0,089 ± 0.008	-	0,010 ± 0,003	0,021 ± 0,002	0,12 ± 0,012
Co-Cr(WBC)	-	0,044 ± 0,003	0,010 ± 0,003	0,015 ± 0,003	0,070 ± 0,005

Inductively Coupled Plasma-Atomic Emission Spectroscopy (ICP-AES)

The mean concentrations of elements released from the Ni-Cr alloy were approximately 8.5 ppb Ni, 1,0 ppb Cr και 1,5 ppb Mo after 28 days. The mean concentrations of elements released from the Co-Cr alloy were approximately 4,2 ppb Co, 1,0 ppb Cr και 2 ppb Mo after 28 days. A mean corrosion rate (total substance loss per day) of 0,070 μg/d.cm² for Co-Cr alloy and 0,12 μg/d.cm² for Ni-Cr alloy were estimated by ICP-AES measurements, respectively, after 28 days. The statistical analysis of the results identified statistically significant differences in corrosion rates of Ni-Cr alloy and Co-Cr alloy ($p<0.05$). The release of metallic elements from the tested alloys into artificial saliva solution is shown in Fig. 4.

DISCUSSION

The results of the present study imply that the *in vitro* corrosion behavior of Co-Cr alloy is better than that of Ni-Cr alloy. A similar conclusion was reached for Co-Cr and Ni-Cr alloys in previous studies. Geis-Gerstorfer J et al.[11] reported that the average substance loss of the tested Co-Cr-Mo alloys after 35 days ranged from 0.43 to 34.9 μg/cm² while the average substance loss of the tested Ni-Cr-Mo alloys after 35 days ranged from 0.65 to 3.261 μg/cm². Co-Cr alloys exhibit better corrosion behavior than Ni-Cr alloys even after repeated use [14].

Passivity is defined as the condition of corrosion resistance due to the inhibition of active dissolution of an alloy by a more or less spontaneous formation of a dense, thin (usu-

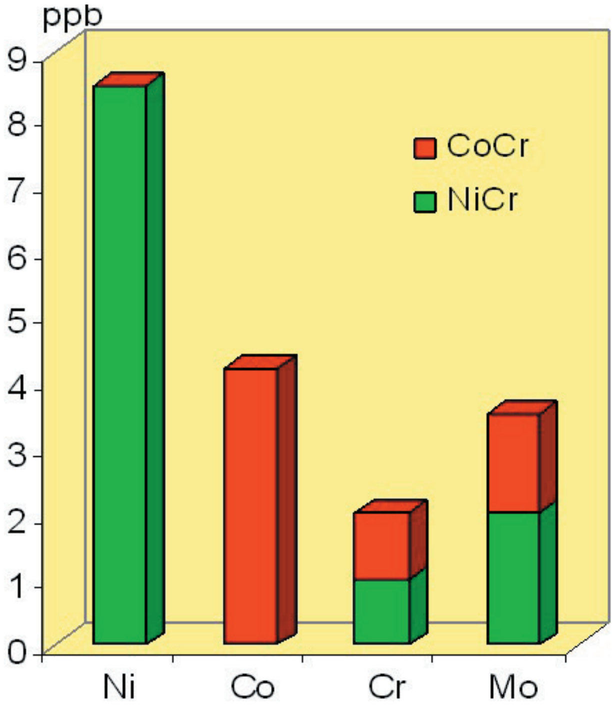

Figure 5. *Release of Ni, Co, Cr, Mo from the test alloys into artificial saliva solution.*

Table 4. *Summarized results from electrochemical (OCP, LP) and analytical (ICP-AES) methods.*

Test Alloy	*E_{corr} vs. time (mV)*	*Mean corrosion rate by Linear Polarization (μg/d.cm^2)*	*Mean corrosion rate by ICP-AES (μg/d.cm^2) Immersion period:28 days*
Ni-Cr	220	0,224	0.12
Co-Cr	280	0,139	0.070

ally <5nm), three-dimensional metal oxide film of limited ionic conductivity[18,19]. In conventional dental materials science, if an alloy is oxidized and the oxide does not break down under physiologic conditions, the alloy is said to be passive or passivated [18,19]**.** The phenomenon of passivity determines the extent of corrosion of Ni-Cr and Co-Cr alloys. A survey of the related literature shows that the corrosion resistance of the alloys can be drastically altered by alloying. This is possible because the composition, structure and properties of the passive films depend on the alloy composition[9,20]. It has been suggested that Ni-Cr alloys containing at least 16-22% Cr and 9-14% Mo can form a complex stable passive oxide layer that provides adequate corrosion resistance[21], whereas a low Cr and Mo content in the Ni-Cr-Mo alloy is associated with a high corrosion rate and susceptibility to accelerated corrosion processes[8]. Reclaru et al [22] showed that the presence of precious metals (Au, Pt, Ru) can deteriorate the corrosion behavior of Co-Cr dental alloys in a significant way. Other researchers have reported that the addition of beryllium (Be) in Ni-Cr dental alloy composition deteriorates their corrosion behavior due to the formation of non-uniform surface oxide film which is more susceptible to breakdown [9,11,23].

From experience in industry it is known that there is a relationship between Cr, Mo and W concentration and the corrosion characteristics. This relationship was specified in DIN 13912 [24]:

$$Cr + 3{,}3(Mo + 0{,}5W) \geq 30$$

The value for an alloy derived from this empirical equation should be above 30 to guarantee adequate corrosion resistance. According to this equation, the values for WBC and W99 are 54,05 and 53.89 respectively, i.e. the value of WBC is higher than that of W99.

For Fe-Ni-Cr-Mo stainless steel, the relative effectiveness of Cr and Mo content on pitting or crevice corrosion usually can be assessed qualitatively by pitting resistance equivalent (PRE), which is represented by the empirical equation [25]:

$$PRE = \%Cr + 3.3\%Mo$$

Geis-Gerstorfer and Greener[10] showed that for a Ni-Cr-Mo dental alloy, a PRE of at least 43 is required to avoid pitting corrosion in the clinically relevant pH range of 4 to 7. Huang [8] reported that a high PRE value of 49 is crucial to protect the Ni-Cr-Mo dental alloy from pitting corrosion in acidic artificial saliva. According to this equation, the values for WBC and W99 are 45.80 and 53.89 respectively, i.e. the value of WBC is lower than that of W99. However, it should be noted that the pitting corrosion equivalent fails to describe the beneficial influence of W on the corrosion resistance of WBC.

Chromium (Cr), Molybdenum (Mo) and Tungsten (W) are alloy elements with strong beneficial influence on the corrosion resistance of Ni-Cr and Co-Cr alloys. The passive film of these alloys consists essentially of Cr in its trivalent state. The presence of an inner oxide layer rich in Cr (barrier layer) is a primary factor in enforcing passivity. Mo is incorporated into the passive film, showing complicated oxide chemistry in several oxidation states, forming aqua and oxo complexes and oxide halides [26]. However, the exact mechanism by which Mo enhances their corrosion resistance is less well understood, and a variety of possible mechanisms have been suggested. It has been postulated that Mo on the alloy surface preferentially locates at local defects, which otherwise would act as dissolution sites, and slows anodic dissolution because of its higher metal-metal bond strength. An alternative hypothesis is that MoO_4^{2-} is formed in the solid state in the exterior regions of the film. This MoO_4^{2-} layer is cation selective, and resists the incorporation of anions such as Cl^- and OH^-. Thus, the MoO_4^{2-} layer reinforces the protectiveness of the high-Cr-containing barrier layer by allowing its growth and prevents the breakdown of the passive film and the subsequent initiation of localized corrosion [27]. W behaves identically to Mo [19] and is predominantly located in the outer regions of the film. It is likely that the enhanced corrosion behavior of WBC may be due to its relatively high Cr content and the W addition.

The results of *in vitro* corrosion testing depends upon instrumentation and experimental variables. Variation in study designs and different electrochemical factors make comparisons between the studies somewhat difficult. Furthermore, the preparation and analytical procedure are technique-sensitive and may be a source of variation. Al-Hiyasat et al. [28], reported that Cr release from the Ni-Cr alloys and Co-Cr alloy used in their study was very low, in most cases below the detection limit, in comparison with the release of Mo and Ni. Other studies have also reported similarly low Cr release in several artificial saliva solutions [29,30,31]. In our study, there is an increased release of Mo, Ni and Co from Ni-Cr and Co-Cr alloy compared to Cr. This finding is in agreement with previous investigations [28-31]. However, it is important to note that the minimum detection limit for Cr in our study is 0.2 ppb whereas the relative minimum detection limit in previous studies was 4 ppb [29] or even 10 ppb [28]. Therefore, the difficulty to quantify Cr release may be attributed to the lower sensitivity of the applied instrumentation.

Natural saliva is an extremely complex system with numerous constituents and variable according to the time of day [32]. Thus, exact duplication is impossible. From the pioneer work of Souder and Sweeney in 1931, no fewer than 60 formulae for artificial saliva solutions have been reported in literature. The large number of artificial saliva formulae reveals the relative difficulty to optimize the simulation of the chemical conditions pertaining in the oral cavity. As it was mentioned by Gal et al.[33], detailed examination of the different formulations showed that all of them were arbitrary compilations of substances, which were more or less similar to an assumed composition of human saliva. Sometimes, complete solutions of the designed products could not be obtained. A formula was very often designed and, then, used by others, with or without modifications, but they rarely made any comments. Hence, it is not a straightforward matter to select a suitable *in vitro* test medium, one that contains inorganic and/or organic molecules, to evaluate the corrosion behavior and elemental release of dental alloys and provide information that is relevant to complex *in vivo* conditions. Moreover, caution is required when the corrosion characteristics of dental alloys are evaluated based on data obtained from the use of different artificial saliva solutions.

In our research work, we used the ISO 6871-I, II artificial saliva formula i.e. 0.1 M NaCl and 0.1 M lactic acid with a pH of 2.3. The ISO standardized formula is easily prepared and remains stable during exposure time. Its pH is lower than that of natural saliva, but within the range of pH encountered during the ingestion of food and under bacterial plaque, which often adheres to dental alloys. Its simple composition reduces the possibility of interference with the ICP-AES analysis for element release. Although the ISO artificial saliva solution contains higher chloride content than that of natural saliva, its use is justified because understanding the corrosion behavior of the alloys under worst-case conditions is important. However, the ISO artificial saliva solution does not contain glycoproteins, phosphate buffer and carbonic acid-bicarbonate buffer, which seem to offer sufficient protection against corrosion[34,35,36]. Most importantly, there is a lack of simulation of the complex intraoral florae, plaque accumulation, and its byproducts, which have proven to have corrosive action.

Wataha and Lockhood[37] reported that the release of elements from dental casting alloys continues through 10 months, reaching a constant rate after approximately 2 months. Moreover, it has been demonstrated that multiphase alloys have long-term release, which might be higher than that during the first weeks, while single-phase alloys have various release patterns with increasing or decreasing rate depending on the element released [38]. In the current study, the 28 day exposure time simulated only 1 month. Therefore, theoretically, longer-term studies are required to better simulate in vivo conditions in which release is expected to continue for extended periods. However, it should be noted that the use of nonagitated, nonreplenished artificial saliva solutions for extended exposure time may, sometimes, erroneously lead to the conclusion that the release rate increased initially and subsequently remained constant because of the saturation of the immersion medium [39].

The OCP is measured under static conditions after a prolonged period of time, when the specimen potential has reached steady-state relative to the reference electrode, as the electrical double layer adjacent to the specimen surface in the given electrolyte has stabilized. After the immersion period (1400 h), a shift in corrosion potential (E_{corr}) towards more anodic potentials with increasing exposure time is clearly observed, indicating enhancement of the passive behavior with aging of the passive film. This increase in corrosion resistance could be due to passive film growth, to a change in the composition or to a change in passive film structure.

A driving force is necessary for electrons to flow between the anodes and the cathodes. This driving force is the difference in potential between the anodic sites (specimen) and cathodic sites (reference electrode). This difference exist because each oxidation or reduction reaction has associated with it a potential determined by the tendency for the reaction to take place spontaneously. The OCP is a quasi-thermodynamic measure,[40] and consequently obtaining corrosion tendencies from the determination of corrosion potentials is necessary but not sufficient to ascertain whether a given dental alloy will suffer corrosion under the given set of the oral milieu.

Inductively Coupled Plasma-Atomic Emission Spectroscopy (ICP-AES) is a sensitive, multi-element analytical method used to identify elements and quantify their concentrations in various media. This method is based on the principle that acid-solubilized specimens of materials exposed to temperatures of 5000 ^{0}K to 10000 ^{0}K emit light of characteristic wavelengths that can be captured by photodetectors and used to determine the elemental concentrations [41]. Inductively Coupled Plasma-Atomic Emission Spectroscopy (ICP-AES) was selected over the usual Atomic Absorption Spectrophotometry (AAS) because it is more sensitive to nickel, cobalt, chromium and molybdenum, the four elements of major concern in this report. An additional advantage of the ICP-AES method of analysis is the substantial reduction in cost, which allows more samples to be analyzed. However, ICP-AES method yields inconclusive evidence about the composition and kinetics of the eluted substances, as well as the reactive state of the leached species. From a biocompatibility standpoint, the oxidation state and chemical form of the released metal significantly influences its absorption, distribution, retention half-life, and excretion. Furthermore, it is imperative to realize that the chemical state of the metal is critical to its mutagenic activity. For example, Cr^{+3} is not a mutagen, but Cr^{+6} is. Nickel (II) ions are weak mutagens, but nickel trisulfide (Ni_2S_3) is highly mutagenic[42] , and is a Ni (III) cation..

The corrosion rates estimated by linear polarization are two-times higher than the respective corrosion rates estimated by ICP-AES. This result is not surprising. ICP-AES is an analytical method by which it is possible to quantify elemental release after a predetermined immersion period. The estimation of the corrosion rates is based on the concentrations of the released elements. It is likely that some of the corrosion products might adhere to the alloy surface and would not be available to be detected by the instrumental analysis of the solutes and thus remain undetectable[43].

The amount of release from any alloy is proportional to the surface area of casting alloys exposed to the oral environment. In the case of dental alloys for metal-ceramic restorations, the alloys are in long-term intimate contact with periodontal tissues, only in the “microenvironment” of the subgingival region. Thus, the exposed area is reduced considerably due to the veneered ceramics. The amounts of elements that are released from Ni-Cr and Co-Cr alloys are far below those taken in as a part of the diet. For example, if an average crown would have 2 to 3 cm^2 of surface area then the amount of Ni released from the W99 alloy (approx. 0.27 μg/day) is far below that eaten (400 μg/day). However, it should be stressed that dietary intake levels of metallic cations have a great deal of uncertainty surrounding them that must be taken into account [44]. Moreover, there is no information that the dietary intake levels of metallic cations themselves have any meaning for long-term biologic safety [42].

The release of metallic elements from dental alloys due to intraoral corrosion processes seems inevitable. Corrosion products of dental casting alloys have been detected in saliva and in gingiva adjacent to these alloys [45]. There is evidence that released metal ions can and do gain access to the body, and these metals may be widely distributed [42]. Currently, any association between release of metal and any metabolic, immunological, or carcinogenic toxicity is conjectural; since cause and effect have never been demonstrated in humans [46]. Thus, corrosion is a necessary but not a sufficient condition for adverse biological effects.

The clinical relevance of *in vitro* corrosion tests has been questioned [39]. In fact, *in vitro* corrosion tests should only be considered useful screening tools [40] . The biological safety of alloys appears to be most related to their corrosion and the resultant release of elemental components. As a result it seems prudent to minimize the biological risks by using alloys that have the lowest corrosion susceptibility. Practitioners should be aware of the corrosion properties of any alloy they use. In particular, practitioner should find alloys that release the least mass, i.e. to be inert and know the complete composition of each alloy used.

CONCLUSIONS

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

1. The *in vitro* corrosion behavior of Co-Cr (WBC) alloy is better than that of Ni-Cr (Wiron 99) alloy. The corrosion resistance of Ni-Cr is approx. 54% of that of Co-Cr and approx. 75% according to immersion test and linear polarization, respectively. Nevertheless, both alloys are adequately corrosion resistant.
2. The *in vitro* corrosion properties of Ni-Cr and Co-Cr alloys are dependent on their chemical composition. W99 contain 22,5% Cr and 9,5% Mo. Its susceptibility to corrosion attack is reduced by its % Cr and %Mo content. The addition of W improves the corrosion properties of base dental casting alloys. The enhanced corrosion behavior of WB C may be due to its slightly higher Cr content and to the W addition.
3. It is evident that both alloys release the elements Ni, Co, Cr, Mo as free cations or complexed ions. The release of elements from dental casting alloys is a reality and it is unlikely that we will discover alloys which release nothing into the body and be completely inert. Further research is needed to establish the levels of release that are at low risk to the body and means to limit release of elements to these low-risk levels.

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