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Corrosion Resistance and Cytotoxicity of Copper-Based and Nickel-Chromium Alloys for Cast Post and Core Fabrication

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A B S T R A C T

Objectives: This study assessed the corrosion resistance and cytotoxicity of two copper-based alloys and one nickel-chromium (Ni-Cr) alloy used for post and core fabrication.

Materials and Methods: In this in vitro study, the corrosion resistance of 9 specimens of dental casting alloys including one Ni-Cr-based (VeraBond) and two copper-based (American Dent-All and Aalbadent NPG) alloys (n=3) was assessed by electrochemical impedance spectroscopy (EIS) in artificial saliva in two different pH levels of 2.5 and 7.1. Their cytotoxicity was evaluated by the methyl thiazolyl tetrazolium (MTT) assay using 15 specimens (n=5). Statistical analysis was performed by one-way ANOVA (alpha=0.05).

Results: EIS showed that the Ni-Cr alloy had the highest corrosion resistance. Aalbadent NPG and American Dent-All alloys showed significantly lower corrosion resistance than the Ni-Cr alloy ($P<0.05$). Aalbadent NPG showed higher corrosion resistance than American Dent-All in early hours, but its corrosion resistance decreased over time and became similar to that of American Dent-All at later time points. Although all groups showed higher corrosion in acidic environment, Ni-Cr showed good corrosion resistance in acidic pH. The cytotoxicity test revealed a significant difference between the copper-based groups ($P<0.05$). Ni-Cr was the most biocompatible alloy amongst all, followed by Aalbadent NPG. American Dent-All showed a high degree of cytotoxicity.

Conclusion: The findings of this study raised some concerns regarding the clinical suitability of copper-based alloys for dental treatments, and the first choice for cast post and core restorations should be Ni-Cr alloys because they are more resistant to corrosion, and are less cytotoxic.

Keywords: Corrosion; Cytotoxicity Tests; Dental Alloys; Dielectric Spectroscopy; Materials Testing

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INTRODUCTION

Endodontic treatment is indicated when irreversible pulpitis is diagnosed, or there is substantial destruction of tooth structure [1]. Depending on the amount of remaining tooth structure, a post and core restoration may be required to provide sufficient retention for the crown [2]. Traditionally, cast metal posts are used to restore such teeth; however, some researchers advocate the use of non-metallic materials [3]. But a high number of researchers have shown that cast metal posts fabricated from custom patterns of the post space have high success

rate in the long-term, and are still preferred by many dental clinicians [4, 5].

Post and core restorations are in contact with the dental tooth structure and gingival tissue. Thus, the corrosion resistance and cytotoxicity of alloys used for the fabrication of such restorations are important parameters to consider. Several reports have linked root fracture to corrosion products of post and core systems [4]. Biocompatibility is also of great significance in this regard since it has been well confirmed that adhesives and cements are not leak-proof, and are dissolved in body fluids over time [5, 6]. Also, root cementum and dentin are permeable to a variety of liquids [5, 6]. Thus, microleakage may occur and result in release of metal ions into the oral environment and periodontium; the subsequent metal-protein or metal-cell interactions may cause cytotoxicity [7]. Many factors such as alloy composition, casting and heat treatment, pH level, and contact between different metals can alter the corrosion resistance and biocompatibility of the alloys [8]; thus, these factors must be taken into account in choosing an alloy for dental restorations.

Noble alloys such as gold in particular have high corrosion resistance and low cytotoxicity; but due to their high cost, their use is limited and base metal alloys are used instead for the majority of cast dental posts [7, 9]. Nickel-chromium (Ni-Cr) is among the most widely used alloys for the fabrication of cast post and core restorations. Many studies have shown high success rate and higher fracture resistance of Ni-Cr post and core systems [10]. Nickel and its corrosion products can cause problems in the oral cavity such as allergy and gingival discoloration; however, it has been well documented that a 17% to 20% chromium content in Ni-Cr alloy creates a stable oxide layer and a strong passivating effect, which makes this alloy highly resistant to corrosion [11, 12]. Also, the long-standing history of successful use of nickel in dentistry, and absence of noticeable reports regarding its adverse biological effects, have proven nickel to be safe for use in the oral cavity [12], and Ni-Cr is still the most widely used alloy for post and core fabrication [13].

In the recent years, a non-precious gold-color alloy (NPG) was introduced to the market, which has a high copper content [9, 14, 15]. It has the mechanical properties of type III gold but at a lower cost, and has a modulus of elasticity close to that of dentin. Some studies have claimed that this material has optimal physical and mechanical properties coupled with much better handling and polishing capabilities [9, 14-18]. The manufacturers claim that it can be used for post and core restorations, and other researchers have studied different aspects of this material for this purpose [9, 14, 15, 17, 18]. Generally, high-copper alloys are severely toxic [11, 19], and corrosion of copper alloys is much higher than that of other alloys [19].

Various in vitro methods may be used to evaluate the corrosion resistance of materials. Electrochemical impedance spectroscopy (EIS) is a reliable quantitative method commonly used for the rapid assessment of protective coatings' corrosion resistance. Also, this procedure is minimally invasive with no forced oxidation or reduction in the open-circuit mode [20, 21].

The methyl thiazolyl tetrazolium (MTT) assay is widely used to assess the cytotoxicity of materials since it is inexpensive, fast, and simple. The MTT salt is converted to blue-magenta colored formazan crystals by the activity of mitochondrial dehydrogenases in living cells. The absorption of dissolved formazan in the visible spectrum correlates with the number of intact living cells [22].

Information regarding the corrosion resistance and cytotoxicity of copper-based alloys is scarce. Thus, it is important to assess their biocompatibility if these alloys are to be used for post and core fabrication. Thus, the aim of this study was to compare the corrosion resistance and cytotoxicity of two different copper-based alloys in comparison with a Ni-Cr alloy.

MATERIALS AND METHODS

Specimen preparation:

The protocol of this in vitro study was approved by the ethics committee of Shiraz University of Medical Sciences

(IR.SUMS.DENTAL.REC.1398.084). One Ni-Cr alloy (VeraBond, Aalbadent, Fairfield, CA, USA) and two copper-based alloys namely Aalbadent NPG (Fairfield, CA, USA) and American Dent-All (Glendale, CA, USA) gold-color alloys were evaluated in this study. Table 1 presents the composition of the alloys.

Nine wax cubes measuring 12×12mm with 3mm thickness were designed and milled from wax discs, which are used for dental casting purposes (YAMAHACHI dental, Japan) using Roland DWX-4W computer-aided design/computer-aided manufacturing machine (ROLAND DGA, Irvine, California), and were cast with the conventional lost-wax technique and phosphate bonded investment material, and poured into a casting ring after vacuum mixing. This process was followed by the wax elimination step using the wax burnout furnace. The alloy ingots were melted under vacuum conditions using an induction casting furnace and later bench-cooled and polished [16]. These specimens were used for the EIS (n=3). The sample size for the EIS was determined based on similar studies [23, 24]. Fifteen other wax cubes measuring 10×10mm with 2mm thickness were also fabricated and cast using the same method, which were used for the MTT assay (n=5). The sample size for the cytotoxicity test was determined by power analysis with an α of 0.05 and a β of 0.80 to detect a difference of 0.30 in loss of cell viability compared to the control (untreated) cells. The power analysis indicated that a minimum of three specimens were needed to attain an 80% power with a 95% confidence interval [25]. The specimens were fabricated using an induction casting machine (Degut Ron eco, Degu dent GmbH, Germany) in accordance with the manufacturer's instructions. After casting, they were polished with blunt instruments, air abrasion, and sandpapers with grit sizes between 200 and 600. They were finally cleaned with sonication in distilled water followed by degreasing with acetone, and were kept in airtight bottles to prevent contamination.

EIS:

Artificial saliva (Nikceram Razi Co., Esfahan, Iran) was used as an electrolyte solution composed of (all units in g/L): KCL=0.964, KSCN=0.189, NaCl=0.126, $K_2H_2PO_4$=0.655, Na_2SO_4=0.337, NH_4Cl=0.178, $CO(NH_2)_2$(urea)=0.2, $CaCl_2.2H_2O$=0.228, $NaHCO_3$=0.630. A neutral solution was prepared with a pH of 7.1 to simulate neutral pH in the oral environment (adjusted by NaHCO3), and an acidic solution was prepared with a pH of 2.5 to simulate an inflammatory condition (adjusted by titration of 1 molar HCl) [26].

EIS was performed by a potentiostat/galvanostat device (PARSTAT, AMETEK, Inc. Berwyn, Pennsylvania, USA) with a three-electrode configuration. Dental alloy specimens were used as the working electrode, platinum as the counter electrode, and Ag/Ag_2Cl_2 with 3M KCl (*E*Ag/AgCl=0.197V/NHE) as the reference electrode. The test specimens were glued to a 50-cc flat cell with an active surface area of 1cm^2, and artificial saliva with two different pH values (7.1 and 2.5) was used as the electrolyte. The test was conducted at 37°C temperature in open-air conditions and 1atm atmospheric pressure. Each specimen was allowed to reach open circuit potential for 1 hour in order to reach a steady state, and the open circuit potential was recorded. The EIS data were recorded in a frequency range of 10MHz to 100kHz and potential range of ±10mV at six time points of 26, 74, 98, 194, 242 and 578 hours (1-24 days) [23, 24, 27-29]. The EIS data were measured using ZView Software (AMETEK, Inc. Berwyn, Pennsylvania, USA), and Nyquist and bode-phase plots were derived from the results.

Table 1. Composition of the alloys

Alloy	Ni	Cu	Al	Cr	Mo	Mn	Si	Others
American Dent-All (copper based)	4%	85%	9%	4%	-	-	2%	
Aalbadent NPG (copper based)	4.3%	80.7%	7.8%	-	-	1.7%	-	Zn 2.7% Fe 3%
VeraBond (Ni-Cr)	77.9%	-	2.9%	12.6%	5%	-	-	Be 1.9%

-:not present

Data were recorded at 5 frequencies/decade. The diameter of the Nyquist (A graphical representation of electrochemical impedance data) loop has a direct relationship with the corrosion resistance of the surface of the sample. The increase of the semicircle diameter (enlargement) indicates that the alloy has good corrosion resistance, and vice versa [23, 24, 29, 30]. The obtained semicircle corresponds to charge transfer resistance in the oxide/electrolyte or metal/oxide layer (R_{cto} and R_{ctI} respectively), and the specimen with the largest or lowest semicircles was considered as the most resistant or the least resistant against corrosion, respectively. In this study, the constant phase element was substituted for ideal capacitive element for a more accurate fit due to the porous and irregular outer layer of oxide film. In this study, the fixed phase components of Double-layer capacitance at the metal/oxide and oxide/electrolyte interface (C_{dli} and C_{dlo} respectively) were used to replace the double layer capacitance and oxide film's capacitance, respectively. Due to different readings and plots derived from the specimens in the same group, this test was not statistically analyzed, and the plot and measurements with the best fit were reported and used for the purpose of comparison of the specimens. All tests were repeated three times to ensure repeatability [23, 24, 29].

Cytotoxicity:

Human gingival fibroblasts [HGF 1-PI 1 (NCBI 165)] were obtained from the cell bank of the Pasteur institute of Iran. The cells were cultured in Dulbecco's modified Eagle's medium (Biowest, Nuaillé, France) supplemented with 15% fetal bovine serum, vancomycin (50µg/mL), ampicillin (20µg/mL) and Fungizone (0.3µg/mL) (all from Sigma-Aldrich, St. Louis, Missouri, USA). All incubations were carried out at 37°C and 95% relative humidity in an air atmosphere containing 5%CO_2, and the culture medium was changed every 3 days.

The samples were sterilized in a hot-air oven at 160°C for 60 minutes, and 7×10^3 cells in 50µL of the culture medium were placed on the test samples on 12-well culture plates (Becton, Dickinson Labware, NJ, USA) and incubated for 4 hours. After ensuring adhesion of the cells, 500µL of the culture medium was added to the wells, and the wells were incubated for 72 hours; then the medium was removed and 400µL of tetrazolium salt [3-(4,5-dimethylthiazol-2-yl)-2, 5-diphenyltetrazolium bromide] (Sigma-Aldrich, St. Louis, Missouri, USA) was added. The plates were incubated for another 4 hours and 400µL of HCl in isopropanol was added to the culture plate for cell lysis. The plates were placed on a shaker for 15 minutes, and 100µL of each well was transferred to a 96-well plate and optical density of each well was read by an ELISA reader (STAT FAX 2100, Palm City, FL, USA) at 570nm wavelength 27, 31]. Cytotoxicity and viability were calculated based on the following formulae:

$$Toxicity\ \% = (1 - \frac{Mean\ OD\ of\ sample}{Mean\ OD\ of\ control}) \times 100$$

$$Viability\% = 100 - Toxicity$$

All assays were repeated five times to ensure reproducibility. Statistical analysis was performed using one-way ANOVA at $P<0.05$ level of significance performed by using a statistical software package (IBM SPSS Statistics for Windows version 22.0; IBM Corp., Armonk, NY, USA).

RESULTS

R_{ctii} and R_{cto} and C_{dli} and C_{dlo} values obtained by EIS at a pH of 7.1, charge transfer resistance at metal/oxide (R_{cti}) and oxide/electrolyte (R_{cto}) interface, solution resistance(R_s), constant phase element related to the double layer at the metal/oxide interface($CPE_{dl,i}$) and oxide/electrolyte interface($CPE_{dl,o}$) are shown in Table 2. Also, changes in R_{cti}, which correlates with the corrosion resistance, are shown in Figure 1. According to the EIS results, among the samples exposed to neutral environment, Ni-Cr alloy showed the highest corrosion resistance in comparison with Aalbadent NPG and American Dent-All. Aalbadent NPG had a higher corrosion resistance than the Ni-Cr alloy in early hours but had much lower resistance later in the experiment. American Dent-All had the least corrosion resistance.

The data obtained by EIS at a pH of 2.5 are shown in Table 3. Also, changes in the charge transfer resistance, which correlates with the corrosion resistance, are shown in Figure 2.

According to the EIS results, among the samples exposed to the acidic environment, Ni-Cr alloy showed the highest corrosion resistance in comparison with American Dent-All and Aalbadent NPG with a significant difference ($P<0.05$).

Aalbadent NPG had a higher corrosion resistance than American Dent-All in early hours but their corrosion resistance became similar over time.

Table 2. Corrosion resistance of dental alloys at a pH of 7.1

Alloy	Time (h)	Rs ($\Omega.cm^2$)	$CPE_{dl,O}$ ($sn.\Omega^{-1}.cm^{-2}$)	$R_{ct,O}$ ($\Omega.cm^2$)	$CPE_{dl,I}$ ($sn.\Omega^{-1}.cm^{-2}$)	$R_{ct,I}$ ($\Omega.cm^2$)
American Dent-All	26	440.7	4.435×10^{-6}	10860	5.349×10^{-6}	$6.199\times10^{+6}$
	74	385.2	5.473×10^{-6}	17653	5.170×10^{-6}	$4.487\times10^{+6}$
	98	632	1.290×10^{-8}	10448	8.265×10^{-7}	$1.983\times10^{+6}$
	194	431.7	-	-	3.013×10^{-6}	$1.896\times10^{+6}$
	242	448.6	-	-	8.388×10^{-6}	$1.062\times10^{+6}$
	578	428.4	-	-	7.612×10^{-6}	$8.514\times10^{+6}$
Aalbadent NPG	26	368.3	-	-	8.515×10^{-9}	$3.273\times10^{+8}$
	74	185.3	-	-	1.330×10^{-8}	$1.825\times10^{+8}$
	98	101.6	-	-	2.635×10^{-8}	$1.770\times10^{+8}$
	194	416.5	3.273×10^{-6}	7110	9.038×10^{-6}	$5.193\times10^{+6}$
	242	452.5	-	-	3.827×10^{-6}	$2.627\times10^{+6}$
	578	291.4	-	-	4.393×10^{-6}	$1.087\times10^{+6}$
VeraBond Ni-Cr	26	204.7	1.476×10^{-7}	14517	1.022×10^{-6}	$6.306\times10^{+7}$
	74	160.7	1.322×10^{-7}	11862	1.666×10^{-6}	$1.653\times10^{+8}$
	98	123.3	3.338×10^{-7}	13868	2.050×10^{-6}	$1.123\times10^{+8}$
	194	95.59	8.845×10^{-7}	8007	2.648×10^{-6}	$9.996\times10^{+7}$
	242	13.54	1.493×10^{-6}	7762	2.731×10^{-6}	$2.133\times10^{+8}$
	578	155.1	1.465×10^{-5}	4346	3.065×10^{-6}	$9.389\times10^{+8}$

Rs: solution resistance; $CPE_{dl,O}$: constant phase element related to the double layer at the oxide/electrolyte interface ; $R_{ct,O}$: Charge transfer resistance at the oxide/electrolyte interface; $CPE_{dl,I}$: constant phase element related to the double layer at the metal/oxide interface; $R_{ct,I}$: Charge transfer resistance at the metal/oxide interface

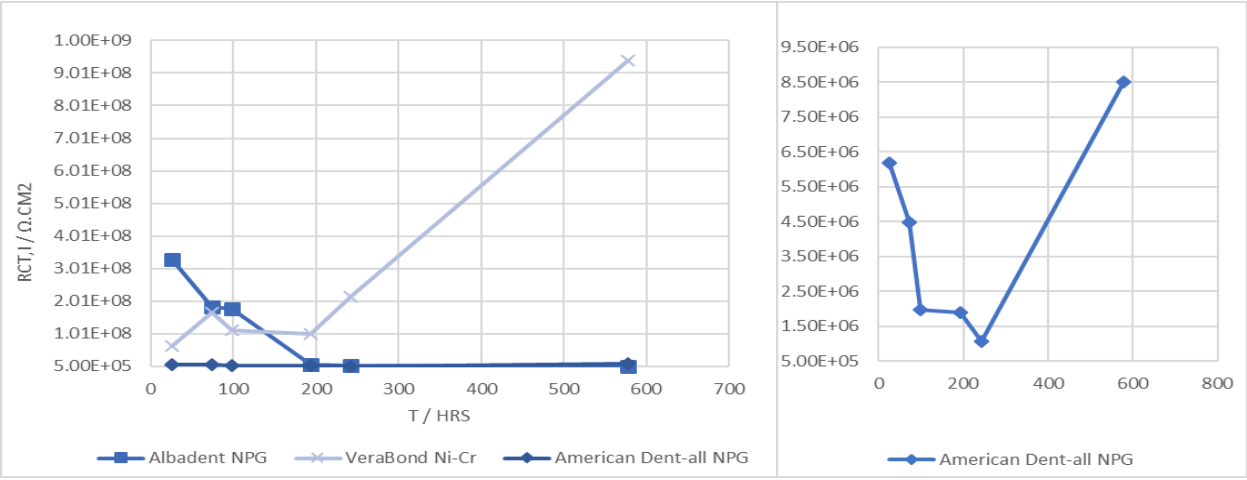

Fig 1. Changes in charge transfer resistance of the metal/oxide layer for 4 different electrodes over time at a pH of 7.1

Table 3. Corrosion resistance of dental alloys at a pH of 2.5

Sample	Time (h)	Rs ($\Omega.cm^2$)	$CPE_{dl,O}$ ($sn.\Omega^{-1}.cm^{-2}$)	$R_{ct,O}$ ($\Omega.cm^2$)	$CPE_{dl,I}$ ($sn.\Omega^{-1}.cm^{-2}$)	$R_{ct,I}$ ($\Omega.cm^2$)
American Dent-All	26	355	1.766×10^{-8}	4868	2.280×10^{-5}	71929
	74	351.25	1.036×10^{-8}	4355	4.218×10^{-5}	75348
	98	108.6	7.139×10^{-9}	4301	4.819×10^{-5}	103170
	194	186	1.327×10^{-8}	4443	5.077×10^{-5}	68651
	242	109.6	1.221×10^{-8}	4907	3.799×10^{-5}	47068
	578	447.5	1.477×10^{-7}	2308	3.620×10^{-5}	34889
Aalbadent NPG	26	202.5	2.004×10^{-8}	14562	1.109×10^{-5}	218230
	74	226.99	3.692×10^{-8}	11014	2.560×10^{-5}	168540
	98	184.94	2.288×10^{-8}	7994	2.599×10^{-5}	82839
	194	98.13	2.454×10^{-8}	8367	3.036×10^{-5}	49260
	242	190	4.294×10^{-8}	3987	1.706×10^{-5}	31901
	578	120.4	3.259×10^{-8}	7863	1.583×10^{-5}	26021
VeraBond Ni-Cr	26	202.2	7.159×10^{-9}	$1.56\times10^{+6}$	2.895×10^{-7}	$5.314\times10^{+6}$
	74	191.6	6.621×10^{-9}	$3.70\times10^{+6}$	2.514×10^{-7}	$5.754\times10^{+6}$
	98	284	6.901×10^{-9}	$1.347\times10^{+6}$	1.957×10^{-7}	$7.441\times10^{+6}$
	194	238.9	8×10^{-9}	$3.309\times10^{+6}$	1.548×10^{-7}	$9.808\times10^{+6}$
	242	318.6	-	-	7.432×10^{-9}	$1.289\times10^{+7}$
	578	210.5	-	-	6.894×10^{-9}	$1.282\times10^{+7}$

Rs: solution resistance; $CPE_{dl,O}$: constant phase element related to the double layer at the oxide/electrolyte interface; $R_{ct,O}$: Charge transfer resistance at the oxide/electrolyte interface; $CPE_{dl,I}$: constant phase element related to the double layer at the metal/oxide interface; $R_{ct,I}$: Charge transfer resistance at the metal/oxide interface

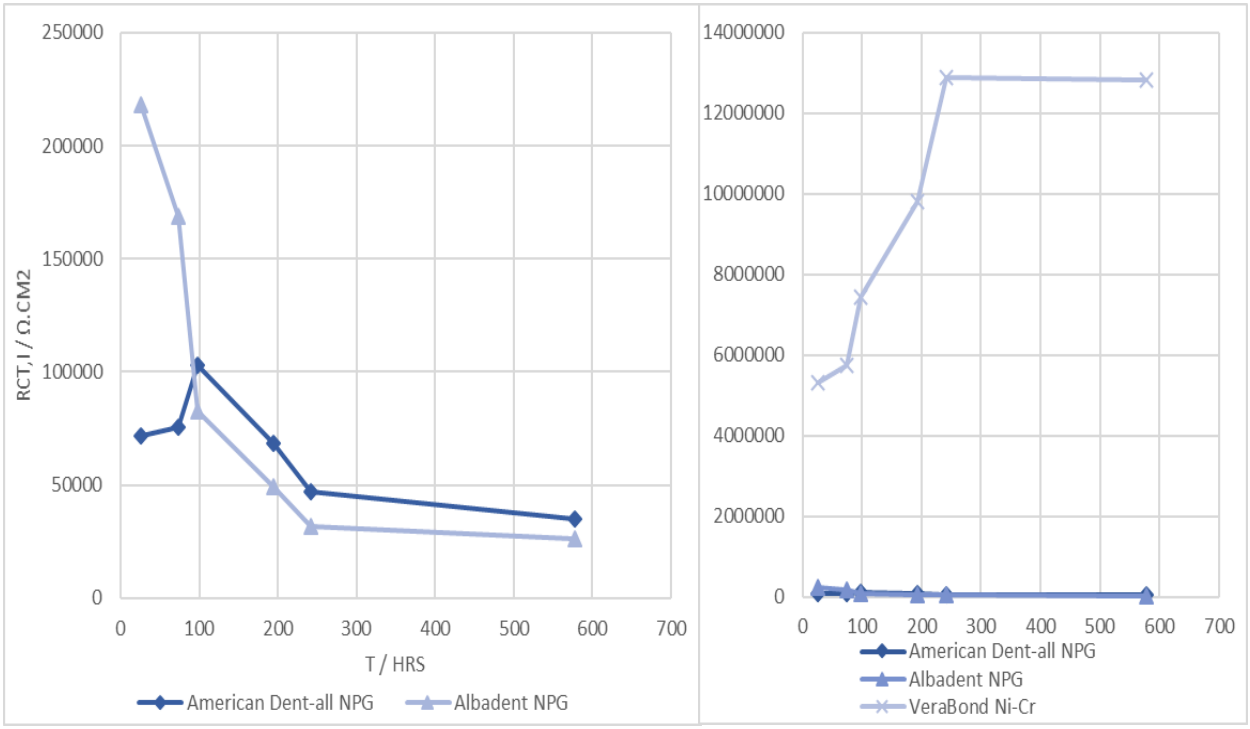

Fig 2. Changes in the charge transfer resistance of the metal/oxide layer for 4 different electrodes over time at a pH of 2.5

Table 4 shows the result of the cytotoxicity assay. One-way ANOVA revealed a significant difference among the American Dent-All alloy, Aalbadent NPG and Ni-Cr group (P<0.05). Ni-Cr was the most biocompatible alloy followed by Aalbadent NPG. American Dent-All was the most cytotoxic alloy as shown in Figure 3.

Table 4. Optical density of different alloys and their cytotoxicity

Alloy	Mean	SD	P value	Cytotoxicity percentage	Viability percentage
American Dent-All (copper)	52.20	14.63	<0.001	86.27	13.73
Aalbadent NPG (copper)	307.400	14.80	<0.001	19.11	80.89
Aalbadent VeraBond (Ni-Cr)	347.00	17.08	<0.001	8.69	91.31
Control	380.00	24.04		0	100

SD: standard deviation

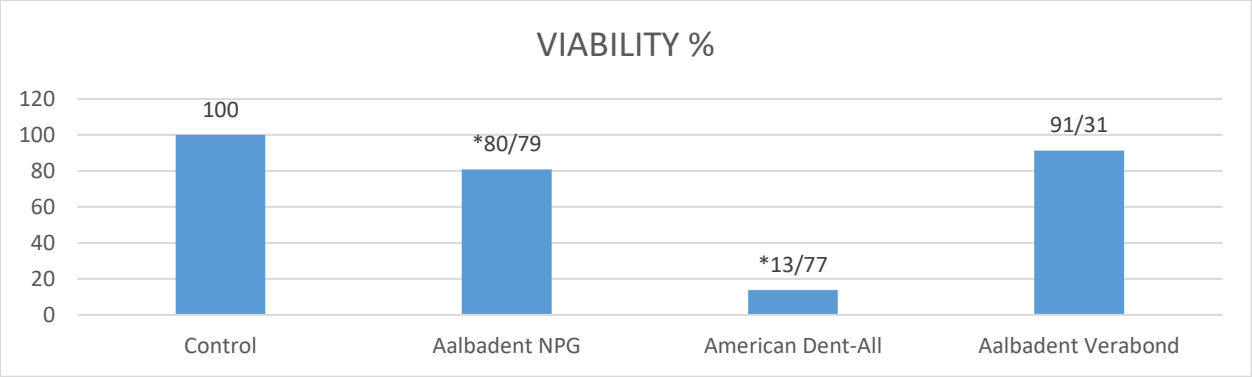

Fig 3. Cytotoxicity of dental alloys according to cell viability. *P<0.05

DISCUSSION

Corrosion of metals in the oral environment takes place continuously because these ions are released through abrasion by foods, liquids, and tooth brushing. Corrosion of dental alloys may result in destructive biological, functional, and esthetic effects. Furthermore, metal ions are released in corrosion processes, and may come into contact with cells and tissues in the oral cavity or disseminate throughout the body. Therefore, corrosion of metallic dental alloys makes them susceptible to failure and consequent problems [32].

In the present study, the properties of Ni-Cr, Aalbadent NPG, and American Dent-All alloys were investigated by using EIS. The results showed that Ni-Cr alloy had the highest corrosion resistance in both acidic and neutral environments, which can be a result of passivation and passive oxide film compounds of Ni-Cr and Mo-Cr-Ni alloys [33].

In an in vitro study, Rao et al. [28] showed that Ni-Cr samples showed high corrosion resistance in neutral, and lower corrosion resistance in acidic environments. This finding was similar to the results of the current study. The reason for the differences in corrosion rate in different electrolytes may be the regular destruction and repair reactions of passive oxide layer in acidic environments [34].

At the initial hours of the test (in neutral pH), the corrosion resistance of Ni-Cr alloy was slightly lower than that of Aalbadent NPG alloy. A probable reason for this finding may be the faster formation of aluminum oxide layer in Aalbadent NPG in comparison with chromium oxide in Ni-Cr alloy [27, 35, 36].

The Aalbadent NPG alloy showed significantly lower corrosion resistance than Ni-Cr but higher than American Dent-All specially in

early hours. Aalbadent NPG electrode has a low Ni and Mn content and has no Mo element. American Dent-All NPG does not contain Ni, Mo or Mn. Mn improves the intergranular corrosion resistance [20]. Such differences in composition may also cause a lower resistance to corrosion in acidic environments [27].

Aalbadent NPG alloy showed superior corrosion resistance in early hours than American Dent-All, but its corrosion resistance decreased over time, which may be due to the difference in protective behavior of their constituents. In general, copper alloys are less corrosion resistant such that Ardlin et al. [27] confirmed that copper-based alloys released 20 times more metal ions into the solution than other types of dental alloys. Moreover, Fateh et al. [37] showed that copper alloys suffered from tarnish and surface alterations; while, the Ni-based group showed less corrosion.

The intermetallic particles containing iron and copper destroy the integrity of the oxide layer in metal surfaces. Copper-based alloys have higher amounts of iron, which results in initiation of pit corrosion and related risks, which was clearly seen in Aalbadent NPG alloys [27, 30].

According to the results of the current study, copper alloys had higher corrosion rate in acidic environment than neutral electrolyte. It could be due to the interactions of copper with thiocyanate and Cl^- ions (chaotropic ions) in acidic environments, which makes a weak complex of copper thiocyanate ($SCN^-+Cu\rightarrow CuSCN$), ($Cu+Cl^-\rightarrow CuCl_2$ and $CuCl$ and $CuCl^-$) that in turn makes copper dissolve at the alloy surface in acidic electrolytes [27, 34, 35]. This was in agreement with the results of Elshahawy et al, [38] who reported that element release increased in lower pH values and resulted in a higher corrosion rate.

According to the results of the present study, Ni-Cr is a biocompatible dental alloy; AalbaDent NPG showed low cytotoxicity (although it was statistically significant) but American Dent-All showed high cytotoxicity. Contrary to the present findings, Liu et al. [39] showed that Aalbadent NPG had severe cytotoxicity. Such differences could be due to different methodologies. They used soluble extract of the samples in order to assess their cytotoxicity while in the present study, cytotoxicity was assessed through direct contact of alloys with fibroblasts.

In the present study, American Dent-All copper alloy showed high cytotoxicity unlike Aalbadent NPG, which could be explained by considering the corrosion rate and its correlation with cytotoxicity as well as the duration of exposure of fibroblasts to the samples. Aalbadent NPG had higher corrosion resistance in the first 74 hours in both acidic and neutral pH environments, which was probably the reason for its low cytotoxicity.

The present study indicated that Ni-Cr alloy is biocompatible due to its higher chromium and molybdenum content. These elements form an insoluble layer which prevents the release of Ni and other metallic ions, and reduces ionic interactions with cells and tissues in the periodontium [33].

American Dent-All copper alloy was the most cytotoxic alloy in the present study. Absence of molybdenum and manganese in the composition of this alloy can be the main reason for its cytotoxicity. In line with the current study, Al-Hiyasat and Darmani [40] showed that if the amount of copper in NPG alloy exceeds 70%, it would be significantly toxic and also the toxicity rate of copper is known to be significantly higher than nickel.

This study had several limitations such as lack of simulation of oral conditions due to its in vitro design. There are indirect MTT tests that are carried out with extract solution containing alloy ions that may provide more accurate results especially when combined with plasma mass spectroscopy. Additional experiments should also be performed to confirm the results of EIS using auger electron spectroscopy and scanning electron microscopy.

There are several types of copper-based casting alloys in the market that are being used without comprehensive information available about them, and the present results cannot be generalized to them. Also, assessment of corrosion resistance and cytotoxicity of materials in vitro cannot completely predict their corrosion behavior and cytotoxicity in vivo, and clinical studies are required to cast a final judgment in this respect.

CONCLUSION

The findings of this study raise some concerns regarding the clinical suitability of copper alloys for dental restorations, and the first choice for cast post and core restorations should be Ni-Cr alloy because it is more resistant to corrosion and less cytotoxic. According to the result of this study, Ni-Cr dental alloy has a higher corrosion resistance and biocompatibility than copper-based alloys. Among copper-based alloys, American Dent-All had the least corrosion resistance and high cytotoxicity. Aalbadent NPG had low cytotoxicity but was not corrosion resistant.

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CONFLICT OF INTEREST STATEMENT

None declared.

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