NiTi superelastic orthodontic archwires with polyamide coating

Article *in* Journal of Materials Science Materials in Medicine · February 2014

DOI: 10.1007/s10856-013-5070-7 · Source: PubMed

5 authors, including:

Luis Alberto Bravo-González
University of Murcia

jose m Manero
Universitat Politècnica de Catalunya

Elisa Rupérez
Universitat Politècnica de Catalunya

F.J. Gil
Universitat Internacional de Catalunya

NiTi SUPERELASTIC ORTHODONTIC ARCHWIRES WITH POLYAMIDE COATING.

L.A. Bravo[1], A. González de Cabañes[2], J. M. Manero[2], E. Rúperez[2], F.J.Gil[2].

[1] Unidad Docente de Ortodoncia. Facultad de Odontología. Universidad de Murcia. Murcia. Spain.

[2] Biomateriales, Biomecànica e Ingeniería de Tejidos. Dept. Ciencia de Materiales e Ingeniería Metalúrgica). Universitat Politècnica de Catalunya. Centre de Recerca Nanoenginyeria. Member of Biomedical Research Networking Center in Bioengineering, Biomaterials and Nanomedicine. CIBER-BBN. Spain.

Corresponding autor:

F.Javier Gil. Dept. Ciencia de Materiales e Ingeniería Metalúrgica). Universitat Politècnica de Catalunya. ETSEIB. Av.Diagonal 647 08028-Barcelona, Spain.

Email:francesc.xavier.gil@upc.edu   Tel. 34 93 4016111

Key words: orthodontic archwires, polyamide coating, tribology, nickel ion release, corrosion

ABSTRACT

Twenty orthodontic archwires with 55.2% Ni and 44.8% Ti (% weight) were subjected to a dipping treatment to coat the NiTi surface by a polyamide polymer. It has been selected a Polyamide 11 (PA11) due to its remarkable long lasting performance. The transformation temperatures as well as the transformation stresses of the NiTi alloy were determined in order to know whether the coating process can alter its properties. The adhesive wear tests have been demonstrated that the wear rates as well as the dynamic friction coefficients μ of polymer coated wires are much lower than metallic wires. The corrosion studies have shown that the use of this polymer, as coating, seals the NiTi surface to prevent corrosion and the release of nickel ions. The average decrease of Ni ions release due to this coating is around 85%.

INTRODUCTION

The interest in aesthetic orthodontic appliances is increasing: lingually placed brackets, aesthetically polymeric archwires, ceramic brackets are different examples. Besides the aesthetic reasons, another concern is that the oral cavity represents a harsh environment for a metallic orthodontic appliance of any kind [1-2]. Corrosion of orthodontic appliances has been thoroughly studied [3-6]. Two main concerns are directly related to the effects of corrosion: biocompatibility and appliance performance [7]. The most important aspect is the interaction that the appliance may have with the patient in terms of absorption of corrosion products and the systemic reactions that may arise. Attention has been placed on Nickel as an element able to induce allergenic reactions and since it is one of the most used constituents of the alloys commonly used in orthodontics such as Stainless Steel, Nickel-Titanium and Copper-Nickel-Titanium [8-17].

The use of a polymer as coating can provide a wide range of specific surface properties to a biomedical device. A basic application is sealing the surface to prevent corrosion and the release of nickel ions. Orthopaedic clamps of NiTi have been coated with plasma polymerized polytetrafluorenthylene (PTFE) for this purpose. PTFE is known for its low friction coefficient; therefore it is suitable as coating on guidewires [18].

Similar or even better corrosion resistance than the one found for PTFE, has been observed for plasma polymerization of hexamethyldisilazane (HMDSN) on a NiTi surface. This coating can withstand 2% strain deformation before failure [19]. Yet the higher elasticity of PTFE and the wide application of that polymer in medicine render PTFE the more attractive polymer coating.

A solution-based elastomeric polycarbonate polyurethane had been applied on vascular stents. The coating with 25-30 μm thickness had been applied to decrease the corrosion from 275 to<13 μm/year. Crosslinked polyurethane-urea films with Ta particles as radiopaque filler were synthesized directly on NiTi stents [20-21]. The siloxane-based polyurethane copolymer was used for dip-coating of stents. The study did not provide corrosion data, but showed highly improved hemocompatibility concerning coagulation and inflammation [22].

Tetraethyleneglycol dimethyl ether, polyethylene terephthalate,… are other coatings used in order to attain even hemocompatibility characteristics or designed to support the adhesion and proliferation of endothelial cells on the surface.

The aim of this work is to obtain and determine the properties (roughness, transformation temperatures and stresses, friction behaviour, corrosion, nickel ion release) of a new NiTi orthodontic archwire coated with polyamide and to compare its properties with the conventional NiTi. This new coating presents aesthetic advantages, avoids corrosion and decreases nickel ion release.

MATERIALS AND METHODS.

Twenty orthodontic archwires with 55.2% Ni and 44.8% Ti in weight have been studied. The archwires were analyzed as received, and ten of these were subjected to a dipping treatment to coat with polyamide PA11 the NiTi surface. Specimens measuring 0.46 mm in diameter and 45 mm in length were cut from the archwires. The treatment was carried out when the NiTi archwires were at 90ºC, after the NiTi archwires were immersed in polyamide for 30 seconds. Polyamide at 90ºC is in liquid phase and coated all the NiTi surfaces.

Roughness

The roughness of the samples was obtained by white light interferometer microscopy (NT 1100 model, Wyko). The dimensions of the explored zone were 124 X 94 μm^2. Six measurements were done for each group of samples. Roughness, spatial and hybrid parameters were obtained with Vision32 software from direct 3D analysis of the images captured. A correction for curvature and tilt of the sample was also applied.

Height (Sa), spacing (Sz) and hybrid (Rsk, Rku) parameters, defined according to ISO 4287-1997 and/or ASME B46.1-1995 were obtained.

Sa is also known as the arithmetical mean deviation of the profile and is the most commonly used parameter in surface roughness. The only spacing parameter studied is Sz, and it represents the mean spacing between peaks. The hybrid parameter Rsk is also known as skewness of the profile, and depending on its sign it tells whether the surface is more porous or spiky. Finally, Rsku is also called kurtosis, and it gives an idea of the spikiness of the profile.

Calorimetry.

The transformation temperatures of the alloy obtained were determined by Differential Scanning Calorimetry (DSC, 2920 modulated DSC, TA Instruments) with a heating ramp of 10ºC/min and a cooling ramp of 2ºC/min. The transformation temperatures and enthalpy determination was made with an appropriate software. M_s (martensite start) and M_f (martensite finish) are respectively the starting and finishing temperatures of martensite formation upon cooling from the austenite phase. Similarly, A_s (austenite start) and A_f (austenite finish) are respectively the starting and finishing temperatures of austenite formation upon heating from martensite phase.

Mechanical tests

Tensile tests were carried out on a servo-hydraulic testing machine (MTS-Bionix 858), working at a cross-head speed of 10mm/min. The five NiTi specimens tested were cylinders of 0.457 mm in diameter and of 45 mm in length. The critical stresses (austenite to stress-induced martensite) were determined from these tests, which were performed in artificial saliva at 37ºC [23]. The chemical composition of the physiological saliva is shown in Table 1.

Friction

Adhesive wear tests were performed in a CSM pin-on-Disk tribometer, in accordance to the ASTM-G99 standard, in salivary medium at 37ºC. The underlying principle on this test could be called wire-on-disk because of its analogy with the pin-on-disk test.

As shown in figure 1, the orthodontic archwires were carefully affixed with cyanoacrylate adhesive in a bakelite holder, without adhesive residues, on the wire surface to be tested. The contact wire plane and the disc were in the longitudinal direction to simulate full-arc contact bracket. The tribometer was immersed inside a polymethylmethacrilate container in which artificial saliva solution was circulated with a temperature control at 37ºC (±1ºC).

An angular velocity of the disks of 0.5236 rad/s and a normal load 10N were used. Ideally, the normal load on the wire should have been around 1N to simulate the load in service (in which typical values range from 0,196 to 0,98 N load). However, 10N were employed to ensure that there was a full contact between both surfaces that could influence the determination of the coefficients of friction.

The dynamic friction coefficients μ (the proportionality constant between the friction force and the normal force) were determined and the wear rates (volume loss) for the orthodontic archwires against the materials commonly used for the brackets (manufactured in 316 Stainless steel and Ti-6Al-4V) were also measured.

As the wear test was being performed, gravimetric measures were controlled in a Sartorius Micro Balance CPA26P, in order to determine the weight loss over time by means of a high-precision set of scales. The sensitivity of these measures was ±0.001 mg. With fixed density, the weight lost can be stated as volume loss.

Corrosion tests

Archwires were sectioned in order to yield corrosion testing samples. A 25 mm length of each archwire was cut with sterile orthodontic pliers and isolated with wax (Paraffin wax Sigma Aldrich) at the interphase between the testing solution and air. The corrosion tests were performed following the ISO-standard 10993-15:2000 “Biological evaluation of medical devices. Part 15: Identification and quantification of degradation products from metals and alloys”.

The tests were carried out with a Voltalab PGZ 301 potentiostat (Radiometer, Copenhagen, Denmark) controlled by Voltamaster 4 software (Radiometer Analytical, Villeurbanne Cedex, France). The testing solution was artificial saliva and artificial saliva with fluoride solution kept at a controlled constant temperature of 37ºC. The reference electrode was an Ag/AgCl/KCl electrode (Eº=0.222V). The auxiliary electrode used was a platinum electrode with a surface of 240 mm^2 (Radiometer Analytical, Villeurbanne, France).

In the corrosion tests we defined the following: (a) *Open circuit potential* (E_{OCP}), the potential of an electrode measured with respect to a reference electrode or another electrode when no current flows to or from the material; (b) *Corrosion Potential* (E_{CORR}), the potential calculated at the intersection where the total oxidation rate is equal to the total reduction rate; and (c) *Corrosion current density* (i_{CORR}), the current divided by the surface of the electrode. It is the size of the anodic component of the current which flows at the corrosion potential E_{CORR}.

The Open Circuit Potential (E_{ocp}) was monitored for three hours in order to allow leveling-off of the value before the polarization resistance test. The Cyclic Voltammetry assay was performed by scanning the potential of the alloy of the sample at 0.25 mV/s with the minimum current set at -1 A and the maximum at +1 A, with a minimum range set at 100 μA between -300 mV and +2000 mV around the OCP value. The Open Circuit Potentials (E_{ocp}), Corrosion Potentials (E_{corr}) and Corrosion Currents (i_{corr}) were recorded for the different samples tested.

Nickel ion release

An ion release experiment was carried out with untreated samples, thermally oxidized samples and samples thermally oxidized and boiled in water for 1 hour. The samples were immersed in 6 ml of saliva artificial at pH=7.4, 37ºC for 30 days. At day 5, saliva was renewed to avoid saturation of the medium.

The Ni released was quantified by Graphite Furnace Atomic Absorption Spectroscopy (GFAAS) using an UNICAM 939 spectrometer at 1, 2, 5, 15 and 30 days. The results were normalized by the real surface area of each sample and were mean values of three measurements.

Statistical analysis

The data were statistically analysed using t-Student tests and one-way ANOVA tables with Tuckey's multiple comparison in order to evaluate statistically significant differences between sample groups. The differences were considered to be significant when p-value < 0.05.

All statistical analyses were performed with Minitab software (Minitab release 13.0).

RESULTS AND DISCUSSION.

For all the roughness parameters, the differences between untreated and coated surfaces are statistically significant ($p<0.05$). Polyamide coating decreases Sa from 21.8 to 13.5 nm (Table 2), and Sz from 373.4 to 303.2 nm. Ssk is slightly negative (-0.5) for the untreated surfaces, and is -0.7 nm for the coated samples. The kurtosis parameter is much higher than 3 for untreated surfaces and close to 3 for coated surfaces.

The transformation temperatures were measured during the first heating and cooling cycle. However, the results at different cooling/heating cycles do not shown differences with significance. R-phase peaks were not observed in the different calorimetric plots; only one peak was detected in each cooling/heating cycle. Table 3 shows the transformation temperatures for as-received Ni-Ti and with polyamide coating. From these results, it can be observed that the coating treatment does not change the transformation temperatures. The heating produced in the coating treatment at 90ºC is insufficient to produce precipitates or grain growth which could change its transformation temperatures [24].

The stress required inducing the formation of stress-induced martensite, $\sigma^{\beta\rightarrow M}$ and the one required for the reverse transformation $\sigma^{M\rightarrow\beta}$ at 20ºC and 37ºC were determined. The transformation stresses ($\beta\leftrightarrow$Stress-Induced Martensite) are shown in Table 4. Similar stresses were required to transform the austenite to stress-induced martensite in the coated archwires because the transformation temperatures do not present changes with significance between the original archwires and coated ones.

It is widely known that these critical stresses increase with increasing test temperature [25]. The superelastic behavior is very similar and produces the elastic deformation of an orthodontic wire and the subsequent release of its elastic energy over a period of time, which increases the correcting forces. Generally, it has become an accepted principle in orthodontics that light continuous forces are desirable to achieve physiologic and controlled tooth movement with minimum pathologic repercussions on the teeth and their surrounding tissues. Orthodontic archwires that can deliver such light forces over long distances would appear to be most useful to clinical professionals during the initial alignment phase of fixed appliance treatment [26].

Frictional force has an important consideration in orthodontic mechanotherapy. The results of the frictional tests performed on the achwires under scrutiny on the two different types of brackets (316L stainless steel and the Ti6Al4V alloy) provided the values for the static and dynamic friction coefficients (μ) as it can be seen in Table 5.

It can be observed that the polymer static and dynamic friction coefficients are lower than beta-Ti and cp-Ti, NiTi and NiTiCu orthodontic archwires in both bracket materials tested. From the results of Table 6, the wear rate of polymeric wires is much lower than metallic wires.

This contribution evaluated static as well as dynamic friction, and the results indicated a lower friction at the archwire-bracket interface when a polymer and Ti-6Al-4V or 316 stainless steel are used in comparison with the other 4 alloy archwires.

Clinically, this means that the net force required for translator movement will be lower for polymer and higher when Ti-based and superelastic alloys (NiTi and NiTiCu) wires are used [27].

The results obtained for the electrochemical corrosion are shown in Table 7. The results of the zero current potential (E_{OCP}) defined as the potential of an electrode measured with respect to a reference electrode or another electrode when no current flows to or from the material, are positive values, for the original NiTi around +470 mV and for the coated NiTi the value increases up to +870 mV. The results showed that polyamide improved the corrosion behavior and the coated orthodontic archwire remains in its immunity range.

The corrosion Potential (E_{CORR}) defined as the potential calculated at the intersection where the total oxidation rate is equal to the total reduction rate. In this case, the potential showed negative values form -10 mV for the original NiTi to -75 mV for the coated archwires.

The corrosion current density (i_{CORR}) is the current divided by the surface of the electrode. It is the size of the anodic component of the current which flows at the corrosion potential E_{CORR}. Since by definition the resulting current is equal to zero at that potential, the cathodic component is of equal size, but of opposite sign. The measured resulting current being zero at the corrosion potential, the corrosion current density i_{CORR} can only be obtained by indirect methods, e.g. by the Tafel equation. The Tafel equation states that a and b are the Tafel proportionality constants for anodic (oxidation) and cathodic (reduction) reactions of a metal. The higher the current density at a given potential, the more prone is the material to corrode. NiTi coated with polyamide had the most passive (-35.1

mA/cm^2) and the original NiTi had the most active (-5.1mA/cm^2) critical current density values.

The difference of the current density at a given potential between NiTi is due to the polyamide coating which inhibits the corrosion on the orthodontic archwires. The decrease of the roughness could produce a decrease on the corrosion behavior but this would even be lower.

For the Ni release tests, the highest Ni release occurs within the first day (figure 2). Polyamide coating produces a decrease the Ni release from 120 ng/l to 20 ng/l in 30 days of immersion. Then, the average decrease of Ni release due to the treatment is around 85%. These values do not represent any problem for the human health. The European Union (EU) has accordingly decreed two directives:

- the Ni release from parts in direct and prolonged contact with the skin must be lower than 0.5 microgram/cm^2/week [28].
- all metallic parts that are inserted into pierced ears and other parts of the human body must not have a Ni release rate greater than 0.2 microgr/cm^2/week [29-30].

Wiltshire and Noble [31] showed that in dentistry, allergies to metals were also frequently reported. In particular, allergy to Ni in females was reported to vary from 9-20%, and in orthodontic patients with pierced ears, 30% were allergic to Ni, Cu and Cr. Bass et al. [32]

were surprised that only few papers reported about patients reacting adversely to Ni-containing dental restorations and there was little evidence that Ni adsorption intra-orally exacerbated existing dermatitis. Nevertheless, there were authorities that advised dentists against using Ni for those patients known to be sensitive to the metal [33].

CONCLUSIONS

The use of a polyamide polymer (PA11), as coating for NiTi orthodontic archwires, presents excellent properties with great potential for the future.

The calorimetric studies have shown that the polymer coating process does not change the transformation temperatures as well as the transformation stresses ($\beta\leftrightarrow$Stress-Induced Martensite) required to transform the austenite to stress-induced martensite. This means that the superelastic behavior is very similar than uncoated archwires.

In comparison with the uncoated NiTi, the adhesive wear tests have been demonstrated that the wear rate of polymeric wires is much lower than metallic wires. Clinically, this means that the net force required for translator movement will be lower than for uncoated archwire.

The corrosion tests as the Ni release tests have shown that the polyamide coating inhibits the corrosion and it produces a decrease in the Ni release ions. This is an important result due to one of the main concerns about orthodontic appliances set in the oral cavity is corrosion. Orthodontic appliances are continually under the corrosive action of saliva, extrinsic fluids coming from food intake or products used for the oral care and hygiene. Furthermore, it is well reported that the Ni ions release could lead to an allergic reaction depending on the severity of the corrosion process.

REFERENCES

1. Andreasen G.F. and Morrow R.E. Laboratory and clinical analysis of Nitinol wire. Am. J. Orthod. 1978, 73: 142-149.

2. Andreasen G.F. A clinical trial of alignment of teeth using a 0.019 inch thermal nitinol wire with a transition temperature range between 31ºC and 45ºC. Am. J. Orthod. 1989, 78: 528-536.

3. Abalos C, Paúl A, Mendoza A, Solano E, Gil FJ. Influence of topographical features on the fluoride corrosion of Ni-Ti orthodontic archwires. J. Mater. Sci. Mat. Med. 2011; 22: 2813-2821.

4. Miura F., Mogi M., Ohura Y and Karibe M. The superelastic japanese NiTi alloy wire for use in orthodontics. Am. J. Orthod. Dentofac. Orthop. 1988; 94: 89-96.

5. Burstone CJ., Qin B. and Morton JY. NiTi archwire a new orthodontic alloy. American Journal of Orthodontics. 1985; 87: 445-452.

6. Walker MP, White RJ, Kula KS. Effect of fluoride prophylactic agents on the mechanical properties of nickel-titanium-based orthodontics wires. Am J Orthod Dentofac Orthop 2005; 127:662-669

7. Kaneko K, Yokoyama K, Moriyama K, Asaoka K, Sakai J. Degradation in performance of orthodontic wires caused by hydrogen absorption during short-term immersion in 2.0% acidulated phosphate fluoride solution. Angle Orthod 2004;74:487-495

8. Yokohama K, Kaneko K, Ogawa T, Moriyama K, Asaoka K, Sakai J. Hydrogen embrittlement of work-hardened Ni-Ti alloy in fluoride solutions. Biomaterials 2005;26:101-108

9. Schiff N, Grosgogeat B, Lissac M, Dalard F. Influence of fluoridated mouthwashes on corrosion resistance of orthodontic wires. Biomaterials 2004;25:4535-4542

10. Ogawa T, Yokoyama K, Asaoka K, Sakai J. Hydrogen absorption behavior of beta-titanium alloy in acid fluoride solutions. Biomaterials 2004;25:2419-2425

11. Kaneko K, Yokoyama K, Moriyama K, Asaoka K, Sakai J, Nagumo M. Delayed fracture of beta titanium orthodontic wire in fluoride aqueous solutions. Biomaterials 2003;24:2113-2120

12. House K, Sernetz F, Dymock D, Sandy JR, Ireland AJ. Corrosion of orthodontic appliances- should we care? Am J Orthod Dentofac Orthop 2008;133:584-592

13. McKay GC, Macnair R, MacDonald C, Grant MH. Interactions of orthopaedic metals with an immortalized rat osteoblast cell line. Biomaterials 1996;17:1339–1344.

14. Kerosuo H, Kullaa A, Kerosuo E, Kanerva L, Hensten-Petterson A. Nickel allergy in adolescents in relation to orthodontic treatment

and piercing of ears. Am J Orthod Dentofac Orthop 1996;109:148–154.

15. Berger-Gorbet M, Broxup B, Rivard C, Yahia L'H. Biocompatibility testing of Ni–Ti screw using immuno histochemistry on sections containing metallic implants. J Biomed Mater Res 1996;32:243–248.

16. Bass JK, Fine H, Cisnero CJ. Nickel hypersensitivity in the prosthodontics patient. Am J Orthod Dentofac Orthop 1993; 103:280–285.

17. Grimsdottir MR, Hensten-Pettersen A, Kulmann A. Proliferation of nickel sensitive human lymphocytes by corrosion products of orthodontic appliances. Biomaterials 1994;15:1157–1160.

18. Villermaux F, Tabrizian M, Yahia L, Czeremuszkin G, Piron DL Corrosion resistance improvement of NiTi osteosynthesis staples by plasma polymerized tetrafluoroethylene coating. Biomed Mater Eng 1996; 6(4): 241-254.

19. Yang MR, Wu SK, DC plasma-polymerized hexamethyldisilazane Coatings of an equiatomic NiTi shape memory alloy. Surf Coat Technol 2000; 127 (2-3): 273-280.

20. Trigwell S, De S, Sharma R, Mazumder MK, Mehta JL, Structural evaluation of radially expandable cardiovascular stents encased in a polyurethane film. J Biomed Mater Res B Appl Biomater 2006; 76B(2): 241-250.

21. Mazumder MM, De S, Trigwell S, Ali N, Mazumder MK, Metha JL, Corrosion resistance of polyurethane-coated nitinol cardiovascular stents. J Biomater Sci Polym Ed. 2003; 14(12): 1351-1362.

22. Tepe G, Schmehl J, Wendel HP, Scaffner S, Heller S, Gianotti M, Clausen CD, Duda SH. Reduced thrombogenicity of nitinol stents- in vitro evaluatiion of different surface modifications and Coatings. Biomaterials 2006; 27(4): 643-650.

23. C. Abalos, A. Paúl, A. Mendoza, E. Solano, F.J. Gil. Influence of topographical features on the fluoride corrosion of Ni-Ti orthodontic archwires. J. Mater. Sci. Mat. Med. 2011; 22: 2813-2821.

24. M. Arciniegas, J. Peña, J.M. Manero, J.C. Paniagua, F.J Gil. Quantum parameters for guiding the design of Ti alloys with shape memory and/or low elastic modulus. Phil Mag 2008; 88: 2529-2548.

25. M. Arciniegas, J.M. Manero, J. Peña, F.J Gil, J.A. Planell. Study of new multifunctional shape memory and low elastic modulus Ni-free Ti alloys. Mater Sci Eng A. 2008; 39: 742-751.

26. C.J. Burstone., B. Qin, J.Y. Morton. NiTi archwire a new orthodontic alloy. American Journal of Orthodontics. 1985; 87: 445-452.

27. J.A. Gurgel, R.M. Pinzan-Vercelino, J. Powers. Mechanical properties of beta-titanium wires. Angle Orthod. 2011 ; 81 : 471-473.

28. [4]. European Parliament and council directive 94/27/EC of 30 June 1994, Off. J. Eur. Commun. 1994; L188: 1–2.

29. Commission directive 2004/96/EC of 27 September 2004, Off. J. Eur. Commun. L301 (2004):51–52.

30. Setcos JC., Babaei-Mahamani A., Di Silvio L., Mijör MA., Wilson NHF. The safety of nickel containing dental alloys. Dent Mater 2006; 22(12): 1163-1168.

31. Wiltshire WA., Noble J., Allergies to dental materials. Vital Autumn 2007; 27:39.

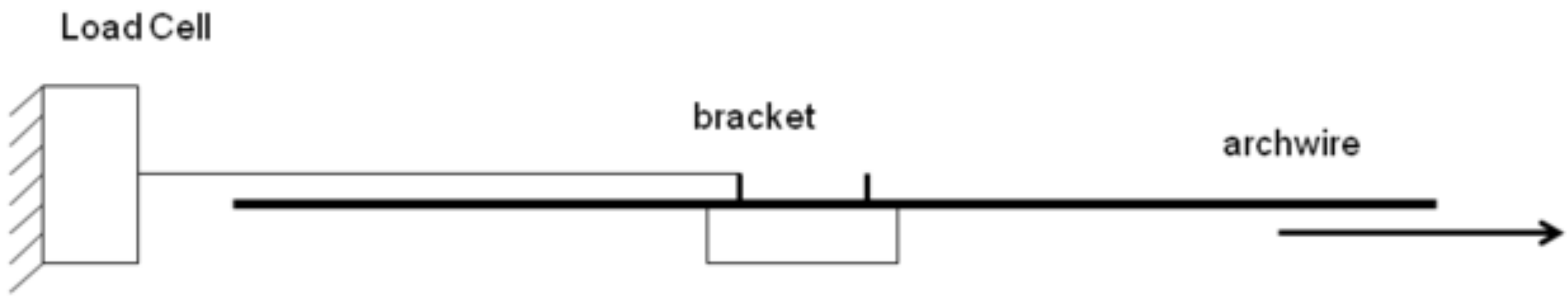
32. Bass JK, Fine H, Cisnero CJ. Nickel hypersensitivity in the prosthodontics patient. Am J Orthod Dentofac Orthop 1993; 103 280–285.

33. Grimsdottir MR, Hensten-Pettersen A, Kulmann A. Proliferation of nickel sensitive human lymphocytes by corrosion products of orthodontic appliances. Biomaterials 1994; 15: 1157–1160.

FIGURE CAPTIONS

Figure 1. Scheme of the test method to determine the friction coefficient.

Figure 2. Ni ion release at different test times. The area of the samples was 1 cm^2.


Load Cell
bracket
archwire

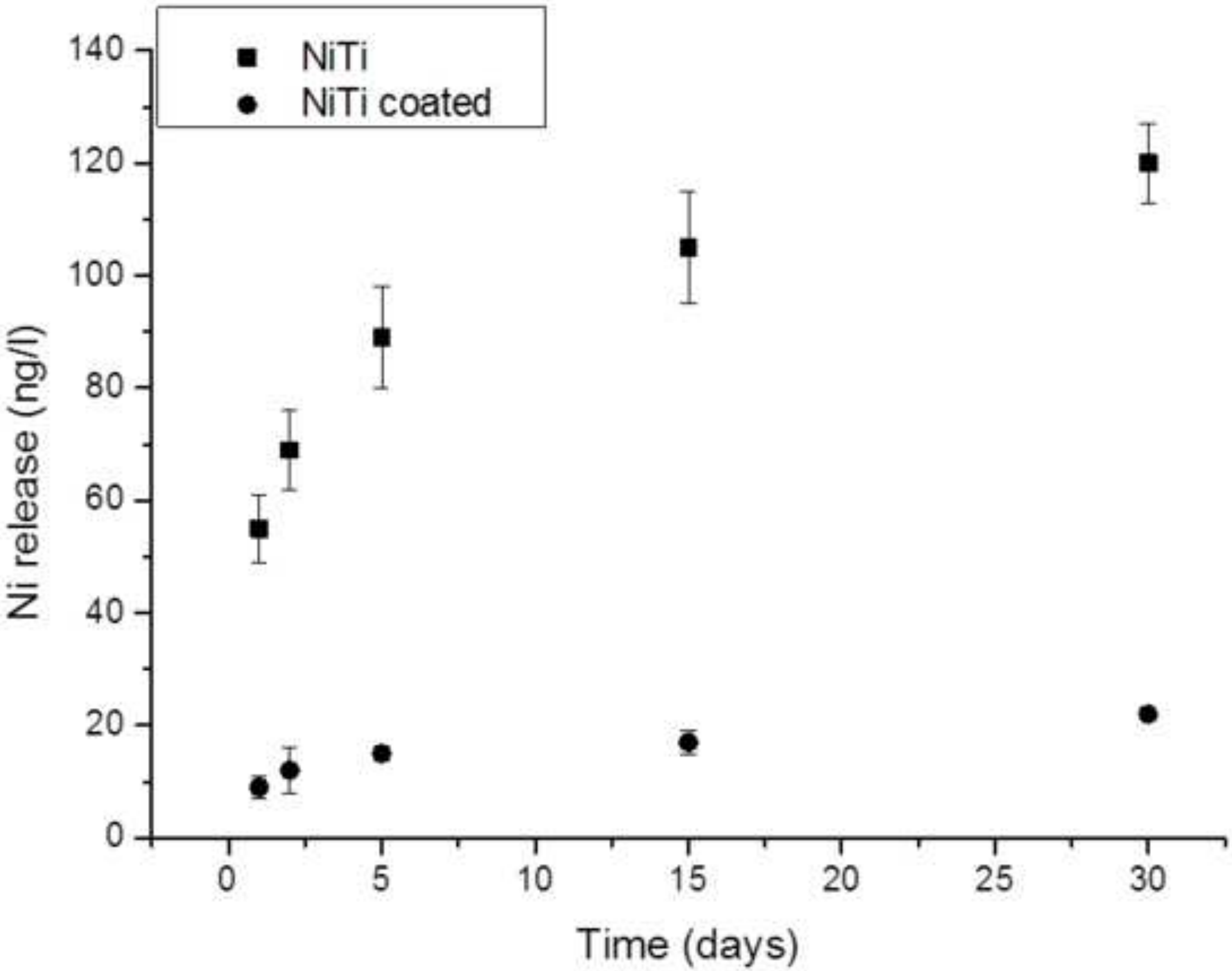

NiTi
NiTi coated
Ni release (ng/l)
Time (days)
0
20
40
60
80
100
120
140
0
5
10
15
20
25
30

TABLE 1. Chemical composition of artificial saliva.

Chemical product	Composition (g/dm^3)
K_2HPO_4	0.20
KCI	1.20
KSCN	0.33
Na_2HPO_4	0.26
NaCl	0.70
$NaHCO_3$	1.50
Urea	1.50
Lactic acid	up to pH=6.7

Table 2. Topographical parameters for untreated and coated samples of NiTi

	Sa (nm)	Sz (nm)	Rsk[b]	Rku[b]
Untreated	21.8 ± 4.8	373.4 ± 58.1	-0.5 ± 0.3	10.0 ± 4.2
Coated	13.5 ± 9.7	303.2 ± 93.3	-0.7 ± 0.1	3.1± 0.2

[a] mean value of ten measurements ± standard deviation

[b]dimensionless

TABLE 3. Transformation temperatures in Celsius for the different NiTi archwires.

Alloy	M_s	M_f	A_s	A_f
NiTi	34.6 ± 0.4	25.8 ± 0.5	33.9 ± 0.4	42.5 ± 0.4
NiTi coated	33.8 ± 0.1	25.4 ± 0.2	31.5 ± 0.2	41.8 ± 0.3

TABLE 4. Critical stresses at different temperatures. The numbers in parentheses are the standard deviation.

Alloy	$\sigma^{\beta\rightarrow SIM}$ (MPa)		$\sigma^{SIM\rightarrow\beta}$ (MPa)	
	20ºC	37ºC	20ºC	37ºC
NiTi	233 (15)	321 (22)	54 (5)	210 (19)
NiTi coated	244 (12)	354 (23)	45 (12)	209 (23)

Table 5. Static and Dynamic friction coefficients of orthodontic archwires obtained against two types of brackets: Ti6Al4V and 316L stainless steel.

STATIC

Archwire / Bracket	NiTi coated	SS ASI304	TiMo	Ti (cp)	NiTi	NiTiCu
Ti-6Al-4V (353 ± 1 HVN)	0,29 ± 0,08	0,36 ± 0,05	0,38± 0,07	0,43 ± 0,11	0,59 ± 0,07	0,49 ± 0,13
SS 316L (435 ± 3 HVN)	0,33 ± 0,10	0,42 ± 0,03	0,45 ± 0,09	0,51 ± 0,13	0,66 ± 0,02	0,57 ± 0,11

DYNAMIC

Archwire / Bracket	NiTi coated	SS ASI304	TiMo	Ti (cp)	NiTi	NiTiCu
Ti-6Al-4V (353 ± 1 HVN)	0,18 ± 0,02	0,26 ± 0,01	0,31 ± 0,02	0,33 ± 0,01	0,51 ± 0,05	0,44 ± 0,03
SS 316L (435 ± 3 HVN)	0,22 ± 0,08	0,32 ± 0,02	0,39 ± 0,02	0,41 ± 0,03	0,56 ± 0,02	0,47 ± 0,01

Table 6. Wear rate of orthodontic archwires (Volume worn in mm^3/h) obtained against two reference materials Ti6Al4V and 316L stainless steel with its hardness values.

Archwire Bracket	NiTi coated	Stainless Steel	TiMo	Ti (cp)	NiTi	NiTiCu
Ti-6Al-4V (353 ± 1 HVN)	0,0027 ± 0,0002	0,0054 ± 0,0001	0,0086 ± 0,0002	0,0102 ± 0,0006	0,0153 ± 0,0008	0,0120 ± 0,0007
SS 316L (435 ± 3 HVN)	0,0033 ± 0,0009	0,0062 ± 0,0009	0,0093 ± 0,0021	0,0113 ± 0,0009	0,0193 ± 0,0008	0,0131 ± 0,0007

Table 7. Mean values obtained from potentiostatic polarization plot of dental materials studied in artificial saliva solution.

Material	E_{OCP} (mV)	E_{CORR} (mV)	i_{CORR} (A/cm2)
NiTi	+478 ± 25	-10 ± 3	-5.1 ± 2.0
NiTi-PA coated	+860 ±34	-75 ± 9	-35.1 ± 3.0