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Laser welding of NiTi orthodontic archwires for selective force application

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Abstract Conventional superelastic orthodontic wires are arch-shaped, have the same mechanical properties all along their length and are used to correct the position of teeth. The disadvantage of these archwires is that there are different types of teeth in the mouth and different forces are therefore needed to rectify their position. The aim of this work was to laser weld several types of NiTi orthodontic wires that had different chemical compositions and superelastic properties, in order to adjust their properties to different parts of the mouth. Microstructural changes, transformation stresses and temperatures, variations in corrosion behaviour and ion release were studied in the welded wires.

Introduction

Tooth movement during orthodontic therapy is achieved by applying forces to teeth, which results in a bone remodelling process. The elastic deformation of an orthodontic wire and the subsequent release of its elastic energy over a period of time give rise to the correcting forces. It is generally assumed that optimal tooth movement is achieved by applying forces that are low in magnitude and continuous in nature [1, 2]. Such forces minimise tissue destruction and produce a relatively constant stress in the periodontal ligament during tooth movement. The superelasticity of nickel-titanium (NiTi) archwires allows the orthodontist to apply an almost continuously low force with larger activation that reduces tissue trauma and patient discomfort, and thus facilitates tooth movement [3, 4].

High-magnitude forces encourage the hyalinisation of the periodontal ligament and may cause irreversible tissue damage such as root resorption. NiTi archwires result in more efficient tooth movement and in a shorter time in comparison to other orthodontic alloys. They are especially appropriate in situations requiring large deflections, such as the preliminary bracket alignment phase in the load deflection stage of orthodontic therapy [5].

The main disadvantage of conventional superelastic nickel-titanium wires is that constant orthodontic forces are exerted on all the teeth. The forces needed to move canines or molars are very different. In this case, the objective of laser welding is to produce multiforce orthodontic wires to optimize the forces applied to different teeth.

The purpose of this study was to investigate the mechanical properties, phase transformation, corrosion resistance and nickel release in artificial saliva at 37 °C of different commercial superelastic nickel-titanium and nickel-titanium-copper orthodontic welded wires.

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Materials and methods

Materials

Three types of orthodontic wires were welded; their composition varied between 48.5 and 50 atomic percentage in

Table 1 Chemical composition in atomic percentage of the archwires studied

Sample	Commercial name	Composition
50TiNi	Nitinol Active Arch–3M Unitek	Ti:50.15% Ni:49.85%
49TiNi	Neosentalloy-GAC	Ti:48.8% Ni:51.2%
CuTiNi	CuTiNi-Cooper 40-ORMCO	Ti:49.29% Ni:45.94% Cu: 4.76%

titanium and their diameter was 0.45 mm. In total, three different samples of a combination of two types of wire were welded. The characteristics of these wires can be seen in Table 1.

Laser welding

The laser used to weld the orthodontic wires was a class-4 Nd:Yag laser working at 1064 nm. The duration of the pulse was 3 ms, the power 2.5 W and the focus zero. The welding was carried out in an argon atmosphere and the welded zone was about 85 μm.

Metallography

The samples were polished metallographically with diamond paste from 5 μm to 0.1 μm and etched with an acidic mixture (17 mL of HF + 33 mL of HNO_3 +50 mL of H_2O). The microstructures were observed using optical and scanning electron microscopes.

Calorimetric tests

The transformation temperatures of the original and welded zones of the different archwires were measured using a DSC. The temperature range of the measurements was from –100 to 100 °C, and the scanning was performed at a rate of 5 °C/min. The transformation temperatures M_s and A_s were measured when there was a sudden increase in the calorimetric signal. In the same way, the final temperatures M_f and A_f were measured when the calorimetric signal returned to the base line [6–8].

Mechanical tests

Tensile tests were carried out on artificial saliva at 37 °C, using a servohydraulic testing machine (MTS-Bionix 858) working at a cross-bar speed of 5 mm/min. The length of the samples was 150 mm. From these tests the critical stresses (austenite to stress-induced martensite) were determined. The chemical composition of the physiological medium is shown in Table 2 [9].

Table 2 Chemical composition of the artificial saliva

Compound	Composition (g/dm^3)
K_2HPO_4	0.20
KCl	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

Corrosion tests

The corrosion tests were controlled using the Voltamaster 4 software and carried out with a Voltalab PGZ 301 potentiostat manufactured by Radiometer Copenhagen. To reproduce corrosion conditions in the mouth, the tests were carried out in an artificial saliva medium at 37 °C, the chemical composition of which is shown in Table 2 [10, 11]. The corrosion cell used was a glass cell with working electrodes and a saturated Ag/AgCl electrode was used as a reference. The test samples were of 10 mm length. The distance from the reference electrode to the samples was 25 mm. The ratio of anode surface to cathode surface was equal to one. The samples were prepared and the corrosion test carried out following the G8 and G15 ASTM standards [10, 12, 13].

The welded archwires were subjected to the following measuring cycle:

- Immersion in the deaerated electrolyte (without oxygen) for 250 min, during which the open potential of each electrode was recorded.
- Using a potentiostat, the potential was increased at intervals of 50 mV up to 1.6 V. Variations in the galvanic current density, potential, etc. were recorded. The polarization curves were recorded in a pseudostationary manner. In our case, 250 min of immersion was sufficient.

Ion release test

Five specimens of each original or welded archwire were placed in a container with artificial saliva. The surface area of the samples was 370 mm^2 and the volume of the fluid was 250 mL. Ten millilitres of the solution was extracted at different times to analyse the metallic ions released. The container was perfectly protected to avoid impurities interfering with the results. The quantification of the ions released was carried out by ICP-MS measurement. This spectrometric technique allows for the quantification of chemical elements in very diluted solutions of even as little as a few parts per billion (nanograms/millilitre). This considerable sensitivity is due to the use of argon plasma, which works at temperatures of between 8000 and 9000 °C, at which almost all materials are ionised. The measurements were taken at 1, 3, 5, 10, 24, 120, 168, 360 and 560 h.

Results and discussion

The transformation temperatures of the original and welded wires are shown in Table 3. The results show that the laser welding of different archwires does not affect the transformation temperatures. The calorimetric peaks overlap due to the transformation corresponding to the two different archwires. The calorimetric curve of the 50TiNi sample and the curves of the samples welded with this archwire are shown in Fig. 1.

Observation by microscopy reveals that the laser welding process does not change the microstructure significantly. The number, morphology and morphometry of the precipitates of the original and welded samples are very similar, even in the welded region. The laser welding leads to melting in a zone of about 6 µm and the heat affects a very small area surrounding it (350 µm). Due to the non-significant variation of the microstructure of the NiTi, the transformation temperatures between the original and the welded archwires are very similar [14, 15]. This study shows that the transformation temperatures remain constant after laser welding. The microstructures of a non-welded wire and two welded wires are showed in Figs. 2 and 3 respectively.

Table 3 Transformation temperatures

Sample	M_s (°C)	M_f (°C)	A_s (°C)	A_f (°C)
50TiNi	49	0	9	71
49TiNi	22	–2	13	48
CuTiNi	19	–10	29	60
50TiNi-49TiNi	50	–3	8	74
50TiNi-CuTiNi	49	–12	8	76
49TiNi-CuTiNi	23	–13	6	63

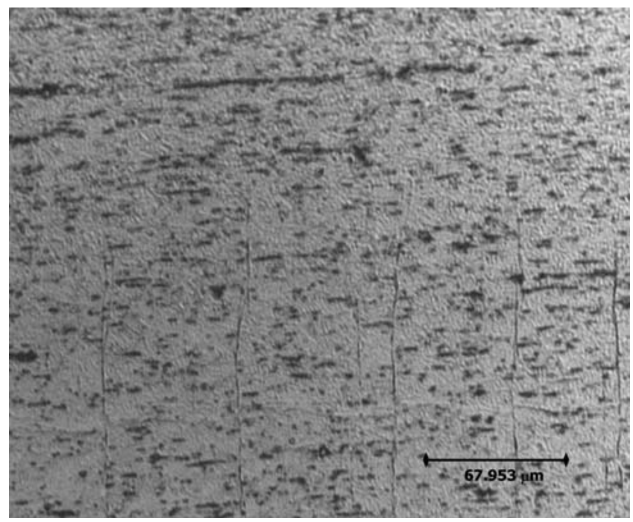

Fig. 2 Microstructure of 50TiNi. The pattern phase is austenite with precipitates TiNi

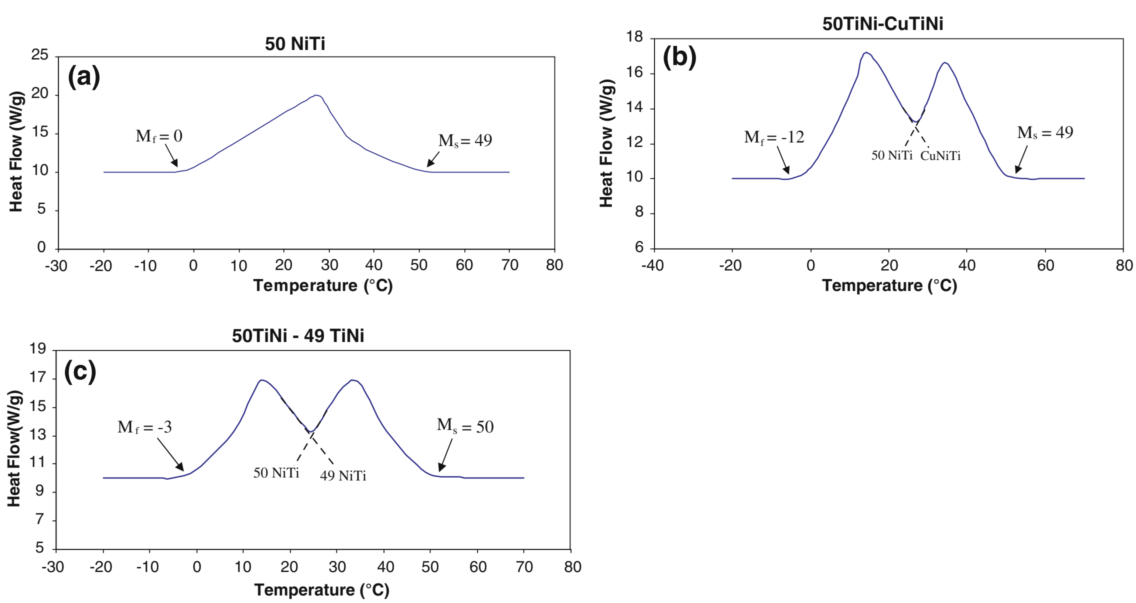

Fig. 1 Calorimetric curves of different archwires: (**a**) 50TiNi archwire. (**b**) 50TiNi welded with CuTiNi. (**c**) 50TiNi welded with 49TiNi

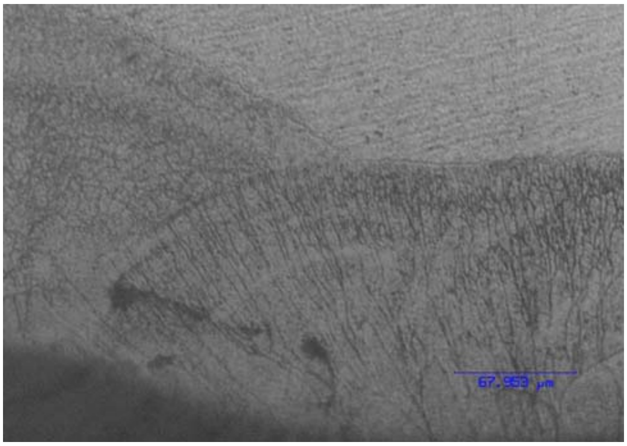

Fig. 3 Microstructure of welding zone. Porous, increase of the number or size of precipitates are not observed

Table 4 Mechanical properties of the wires, (being σ_{max} the maximum strength before fracture and A is the deformation of wires)

Sample	σ_{max} (MPa)	A (%)
50TiNi	1147.8	21.2
49TiNi	1477.6	27.9
CuTiNi	1036.6	21.6
50TiNi-49TiNi	901.2	14.1
50TiNi-CuTiNi	725.0	9.8
49TiNi-CuTiNi	685.3	10.1

The black line in Fig. 2 matches the line created by the pulse of the laser beam. These are stress lines created by a laser pulse. The stress lines lead to a decrease in the maximum strength and the ductility of the material. Therefore, in the tensile tests, failure is always expected to occur at the welding point.

The results of the mechanical tests are shown in Table 4. For the welded archwires, the loss of maximum strength is around 35% and the loss of deformation is 50%. In spite of these losses, the mechanical properties obtained meet the requirements for orthodontic wires in a wide range of applications [15, 16].

A representative stress-strain curve for welded wires at 37 °C is shown in Fig. 4. Based on these results, the

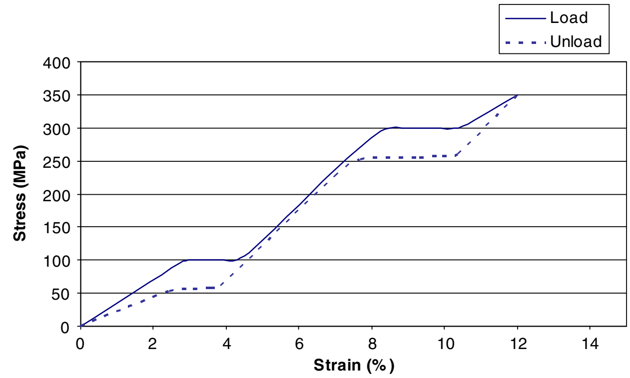

Fig. 4 Stress-Strain curve of 50TiNi-CuTiNi welded archwire. The plateaux correspond to the stress-induced martensite transformation of the different archwires welded

transformation stresses of the welded NiTi archwires are two (austenite $\rightarrow$ stress-induced martensite) and two retransformation stresses (stress-induced martensite $\rightarrow$ austenite), as shown in Table 5. Consequently, laser welding does not significantly affect the superelasticity of the NiTi wires. The stress-induced transformation of both alloys occurs at different stress values; therefore, the welded NiTi wires produce different superelastic stresses in different parts of the archwire [16].

The low corrosion rates of alloys used for dental applications mean that these rates can be easily determined by electrochemical methods. These methods also enable corrosion to be measured repeatedly in the same sample. Table 6 shows the results of the different corrosion magnitudes, which have been adapted to the internationally

Table 6 Mean values obtained from potentiostatic polarization plot of dental materials studied in artificial saliva solution

Material	E_{OCP} (mV)	E_{break} (mV)
50TiNi	–20 ± 5	810 ± 15
49TiNi	–24 ± 7	793 ± 12
CuTiNi	–56 ± 2	825 ± 15
50TiNi-49TiNi	–85 ± 11	785 ± 16
50TiNi-CuTiNi	–105 ± 5	770 ± 17
49TiNi-CuTiNi	–250 ± 8	750 ± 10

Table 5 Transformation stresses

Sample	$\sigma^{A \rightarrow SIM}$ (MPa)	$\sigma^{SIM \rightarrow A}$ (MPa)	$\sigma^{A \rightarrow SIM}$ (MPa)	$\sigma^{SIM \rightarrow A}$ (MPa)
50TiNi	302	295		
49TiNi	323	312		
CuTiNi	119	114		
50TiNi-49TiNi	298	280	333	320
50TiNi-CuTiNi	112	105	300	285
49TiNi-CuTiNi	110	98	330	315

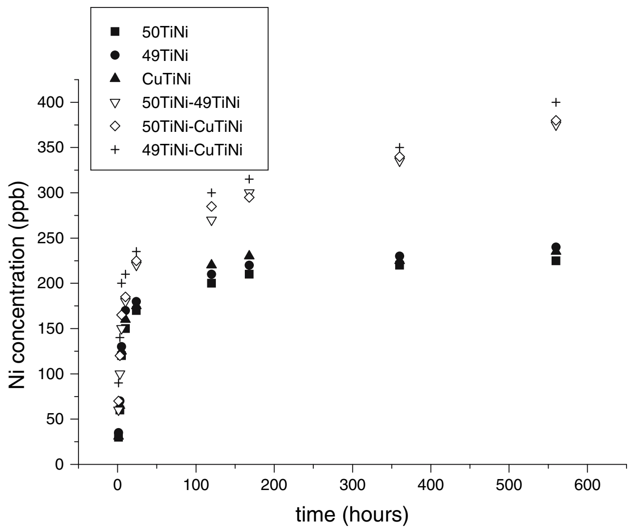

Fig. 5 Nickel release in saliva medium at different immersion times for the different archwires studied. The wires welded increase the nickel release in relation to the original samples

acknowledged definitions used in ISO/CD 10271 ("Dental Metallic Materials-Corrosion Test Methods") [17]. The free potential tests showed that, in all cases, the wires have high corrosion resistance [18, 19]. Although several other differences between the TiNi and the CuTiNi were observed, the main difference is that the addition of copper produced a decrease in the corrosion resistance. It also narrowed the stress hysteresis and stabilized the superelastic characteristics against cyclic deformation, which produced a decrease in the slope of the load-deflection unloading curve with respect to that of the NiTi [20, 21]. Moreover, it led to greater stability in both the transformation temperatures and the forces applied to the teeth [21]. For the welded wires the free potential was lower due to the galvanic couple, which favours corrosion. The cyclic voltammetry results indicate that the passivated layer broke at around 0.8 V for the original archwires and at 0.7 V for the welded ones. The passivated film of the welded archwires broke at low potential.

The ion release tests in the artificial saliva medium showed that the nickel release in the original orthodontic archwires remained under 250 ng/mL after 25 days, and it was lower than the nickel release in welded orthodontic archwires, as can be seen in Fig. 5. To give an idea of the biological significance of this, we can say that this release is lower than the amount of ions contained in drinking water (20 ng/mL of Ni, 15 ng/mL of Cu and 50 ng/mL of Ti) [22]. Therefore, although systematic biocompatibilty tests must be carried out, it is reasonable to assume that the amount of ions released could be within the range of biological tolerance in the use of this alloy as a biomaterial for orthodontic archwires [23, 24].

This paper is an approximation to what happens in the mouth, and therefore further research is needed into the long-term effect of ions on the tissues and organs. Although nickel ions are released into the environment on a small scale, they can accumulate during the useful life of the orthodontic archwire and affect the patient's saliva.

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