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Changes in the mechanical properties of two nickel-titanium archwires after 3 months of clinical usage

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Background: Nickel-titanium archwires have unique mechanical properties that make them the archwire of choice during the first phase of orthodontic treatment. However, during its clinical use when subjected to oral conditions, these properties can undergo great changes.
Materials and methods: A sample of 24 randomly chosen superelastic NiTi orthodontic archwires (12 TE and 12 PSE) with a 0.014-inch round section from the same manufacturer were distributed into four groups of six archwires each. The first two groups were new wires (as-received), which were used as controls (T0), and the other two were collected after 3 months of clinical usage (as-retrieved) in orthodontic patients (T1). Mechanical properties were measured by mechanical tensile testing and three-point bending tests under the same experimental and temperature conditions (36°C) in a universal testing machine. Comparisons between the groups at T0 and T1 were performed with *t*-tests and Mann-Whitney *U* tests. A paired *t*-test and Wilcoxon signed rank sum test were used for intragroup comparisons (T1-T0).
Results: At baseline, PSE wires presented significantly ($P < 0.05$) higher load at fracture, range, ultimate tensile strength (UTS), yield strength, springback, maximum tension and flexural ultimate strength (FUS) than those of TE wires. At T1, a significant decrease in load at fracture and UTS in PSE wires and in FUS in TE wires was found. After 3 months of clinical usage, the changes (T1-T0) in the mechanical properties of both alloys were similar.
Conclusions: After 3 months of clinical usage, wires lost some of their mechanical properties and had less resistance to breakage. However, the as-received differences between both wires were maintained after clinical usage.

1. Introduction

Currently, nickel-titanium (NiTi) alloys are routinely used in orthodontics during the early stages of treatment because their properties of low modulus of elasticity, high springback, and low-force delivery meet the criteria of light and continuous force that are desirable at this stage [1]. Superelastic NiTi (SE) archwires are characterized for their properties of shape memory effect, and superelasticity. Two types of superelastic NiTi alloys are currently available: austenitic active NiTi called pseudoelastic (PSE), these wires change from the initial austenite structure to the martensite structure on loading, with the reverse transformation on unloading; and the shape memory NiTi orthodontic wires are martensite-active alloy called thermoelastic (TE), and change from the initial lower-temperature martensite structure to the austenite structure when transferred from the room temperature orthodontic-office environment to the higher-temperature oral environment [2].

Diverse authors have observed differences in some mechanical properties between PSE and TE NiTi wires [3,4]. Gatto et al. [3] observed that TE NiTi wires exerted significantly lower working forces than PSE wires of the same size in all experimental tests. Gravina et al. [4] found that TE wires presented lower resistance to tension, lower yield strength, higher percentage of deformation on

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constant baselines of deactivation, and slighter loadings of deactivation in relation to PSE.

Different commercially available SE NiTi alloys are offered from a large number of manufacturers that show different mechanical properties that can affect their clinical performance [5–7]. However, it is not clear for the clinician how to choose the ideal NiTi SE wire, depending on the achievement of the objectives for the most efficient tooth movement in a specific malocclusion. In contrast, the final decision of the clinician normally is due to the strong advertising of the manufacturers, their availability at the local markets, and cost.

The oral environment can affect both the surface and mechanical properties of orthodontic wires [8–13]. Bavikati [8] observed a significant decrease in the tensile strength after the clinical usage and autoclave sterilization of NiTi SE wires.

Numerous studies have evaluated the surface properties of NiTi SE archwires after clinical usage, but only a few have assessed the mechanical properties of these wires [8,11,14–16]. The mechanical properties of alloys, such as the ultimate tensile strength (UTS), elastic modulus (E) yield strength (YS), bending load and unloading forces, are affected by different oral conditions, and these changes may reduce their clinical efficiency [13,17]. The aim of the present study was to compare the changes in the mechanical properties of TE and PSE nickel-titanium (NiTi) orthodontic archwires after 3 months of clinical use.

2. Materials and methods

From a total sample of 100 lower NiTi SE archwires (50 TE and 50 PSE) with a 0.014-inch round section from the same arch form (Orthoform III), batch, and manufacturer (3M Unitek, Irwindale, CA), 12 TE and 12 PSE NiTi archwires were randomly chosen for the experiment. The archwires were divided into two categories: new archwires (T0, as-received) and archwires for intraoral use (T1, as-retrieved) during 3 months after bracket placement, and distributed in four groups of six archwires each (T0 = 6 TE and 6 PSE; T1 = 6 TE and 6 PSE).

For in vitro testing and for clinical use, the archwires were blinded to the practitioners, researchers, and machine operators. The wires were tested under the same standardized testing conditions, according the International Norm ISO15841:2006 that specifies requirements and test methods for wires to be used in fixed and removable orthodontic appliances. This norm also refers that sampling six specimens of a single product from one batch shall be procured for each test [18].

The archwires used in vivo were collected from patients treated with fixed orthodontic devices at the lower arch from a randomized clinical trial at the Orthodontics Department at the UniCIEO University (Bogotá, Colombia). The patients provided written informed consent for their use. The patients included in the experiment were recruited based on the following criteria: age range from 13 to 30 years, complete permanent dentition at the lower arch, and moderate lower anterior crowding (4–6 mm) according to Little's irregularity index [19]. Patients with caries, periodontal problems, those who require extraction treatment or need interproximal stripping, intermaxillary elastics or coil springs during the first 3 months of orthodontic treatment at the lower arch were excluded. All the patients used the MBT prescription brackets with 0.022 × 0.028-inch slot from the same manufacturer (Gemini brackets; 3M Unitek).

The archwires retrieved were disinfected with a three-enzyme soap for 5 minutes, washed with water, and dried with tissue paper to avoid any bending. The specimens were identified with a code indicating their type and date of recollection. The wires were stored in plastic bags at room temperature until testing. The new archwires (as-received) were removed from the original manufacturing package only immediately before laboratory testing.

The mechanical properties of the archwires were measured with a universal mechanical tester (AG-5 KN; Shimadzu, Kyoto, Japan) with the assistance of a qualified engineer. The temperature was controlled by a chamber (TCE-N300-CE; Shimadzu, Kyoto, Japan). The temperature was set at 36°C, and the relative humidity registered was 58.5% and 55.7% for the tensile and bending tests, respectively.

2.1. Tensile test

The archwires for mechanical tests were cut in sections of 30 mm over a standard lower arch template in the anterior segment. The specimens were fixed into the jaws of the testing machine, and the distance between the crosshead was standardized as 20 mm. The machine was operated in tensile mode with a crosshead speed of 1.5 mm/min. Load at fracture (N), range (mm), ultimate tensile strength (UTS, N/mm^2), modulus of elasticity (E, N/mm^2), yield strength (YS, N/mm^2) at 0.2% deformation and the maximum elastic strain (springback, YS/E ratio) were calculated, and stress/strain diagrams were obtained.

2.2. The bending test

A three-point bending test was performed in the same universal machine. The center of each wire was deflected at a crosshead speed of 7.5 mm/min under the pressure from a metal pole of 0.36 mm in diameter, and the distance between the crosshead was standardized as 12 mm. Each sample was loaded until a deflection of 3 mm was produced. Maximum tension (N), flexural range (%), flexural ultimate strength (N/mm^2) and stiffness (slope of plot for initial linear range) (N/mm) were calculated.

2.3. Statistical analysis

For statistical analyses, STATA14 software (version 14; StataCorp, College Station, TX) was used. Data normality distribution was assessed with the Shapiro-Wilk test and Q-Q plots. According the data distribution, *t*-tests and Mann-Whitney *U* tests were applied for comparisons between the TE and PSE groups at baseline (T0) and after 3 months of clinical usage (T1) for tensile and three-point bending tests. Comparisons of the mechanical properties between as-received (T0) and as-retrieved (T1) archwires in each group were performed by paired samples *t*-test and Wilcoxon signed rank sum test. The significance level was established at $P < 0.05$. The groups were blinded for the analysis.

3. Results

Table 1 summarizes the results of tensile and bending tests for the PSE and TE groups at T0 and T1 and the differences between them. At baseline (T0) in the tensile test, the TE and PSE archwires presented statistically significant differences ($P < 0.05$) for all the measured mechanical properties except for the modulus of elasticity (E). In the bending test, there were significant differences ($P < 0.05$) in maximum tension and flexural ultimate strength. Meanwhile, at T1 (as-retrieved), statistically significant differences ($P < 0.01$) were observed between TE and PSE NiTi archwires for load at fracture, UTS, YS, maximum tension, and flexural ultimate strength. Regarding the changes after clinical usage (T1-T0), for the PSE wires, the differences were significant ($P < 0.05$) for load at fracture (T1-T0 = 3.63) and UTS (T1-T0 = 35.74). For the TE wires (T1-T0), the only significant difference ($P < 0.05$) was for flexural ultimate strength (T1-T0 = 191.31). For the comparison of the changes after clinical usage between TE and PSE wires, no statistically significant differences were found for any of the studied properties.

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Table 1
Changes in mechanical properties in relation to the time intervals

Properties	T0				T1			
	Wire				Wire			
	PSE	TE	≠	*P*	PSE	TE	≠	*P*
	Ẋ (SD)	Ẋ (SD)			Ẋ (SD)	Ẋ (SD)		
Tensile test								
Break resistence (N)	154.33 (1.11)	120.73 (3.08)	33.61	0.0039** [a]	150.71 (2.52)	124.32 (10.91)	26.41	0.0039** [a]
Range (mm)	5.67 (0.65)	4,88 (0.30)	0.79	0.0231* [b]	5.38 (0.646)	5.13 (1.72)	0.24	0.2002[b]
Ultimate tensile strength (N/mm2)	1516.21 (10.9)	1186.22 (30.3)	330	0.0039** [a]	1480.42 (24.71)	1220.92 (107)	259.53	0.0039** [a]
Modulus elasticity (N/mm^2) (E)	20,344 (1480)	18,483 (4645)	1861	0.386[a]	18,691.23 (5234)	24,086.95 (1206)	5396	0.5218[a]
Yield strength at 0.2% (N/mm^2) (YS)	446.09 (23.23)	316.53 (10.71)	129.56	0.039* [a]	456.41 (49.81)	360.91 (46.21)	95.42	0.0063** [a]
Maximum Elastic Strain (Springback = YS:E)	0.0221 (0.03)	0.0178 (0.03)	0.0043	0.0341* [a]	0.0266 (0.009)	0.0174 (0.006)	0.0092	0.0878[a]
Bending test								
Maximum tension (N)	1.57 (0.39)	1.18 (0.12)	0.39	0.0301* [b]	1.50 (0.07)	1.27 (0.08)	0.22	0.005** [b]
Flexural range (%)	2.30 (0.51)	2.20 (0.77)	0.09	0.8079[a]	1.93 (0.26)	1.49 (0.70)	0.44	0.183[a]
Flexural ultimate strength (N/mm^2)	858.21 (217)	644.41 (67.3)	213.61	0.025* [a]	985.91 (45.91)	835.75 (56.51)	150.11	0.0005*** [a]
Flexural stiffness (N/mm)	3.55 (1.54)	5.53 (2.11)	−1.99	0.0932[a]	4.45 (0.43)	5.56 (1.76)	−1.11	0.1871[a]

PSE, pseudoelastic; SD, standard deviation; TE, thermoelastic; T0, before clinical use; T1, 3 months of clinical use.
Statistically significant: $*P < 0.05$; $**P < 0.01$; $***P < 0.001$; $****P < 0.0001$.
[a] *t*-test.
[b] Mann-Whitney *U* test.

Figure 1A and B shows the stress/strain graphics for PSE and TE wires at T0 from the tension tests, where the martensitic plateau at the TE wires occurred with lower deformation of the wire at lower force and wider martensitic phase than PSE wires. At T1, TE wires showed a narrower martensitic plateau as PSE wires at T0 (Fig. 2A and B).

At the bending tests (Fig. 3A and B), the stress/strain graphics for PSE and TE wires at T0 showed that PSE wires delivered higher force at bending that TE wires. At T1 the same behavior that occurred at T0 was noted with higher forces at bending than T0 for both of the wires (Fig. 4A and B).

4. Discussion

Mechanical properties in orthodontic wires are important for clinical efficiency and to achieve treatment goals. For instance, the UTS measures the wire's fracture strength, the modulus of elasticity describes the resistance to elastic deformation and determines the magnitude of force delivered by a wire activated within the elastic range, the YS is a measure of the point beyond which permanent deformation will occur if the force is increased, and the YS:E ratio indicates the clinical performance of wires, which is indicative of load-deflection curve, working range, stiffness, and resilience [20–22].

In the present study, we found that at baseline (as-received) in the tensile test, PSE wires had higher values in all the measured mechanical properties than TE wires, with statistically significant differences ($P < 0.05$) except for the modulus of elasticity. For the bending test at baseline, PSE wires showed statistically higher values for maximum tension and flexural ultimate strength, confirming the results of the tension test. These results demonstrated that PSE wires are stiffer, with greater fracture strength and better

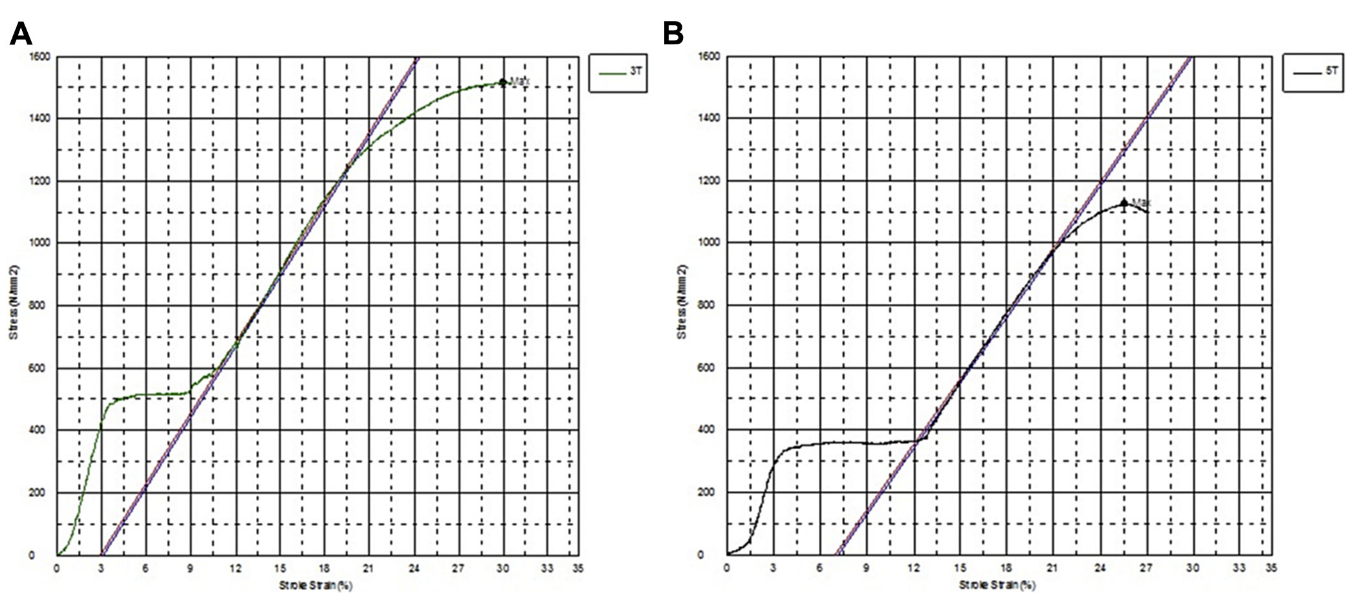

Fig. 1. Stress/strain curves in tensile test at T0. (A) Thermoelastic NiTi wire. (B) Pseudoelastic NiTi wire. The red line in the figures determines the slopes of the plots for continued loading beyond the pseudoelastic region.

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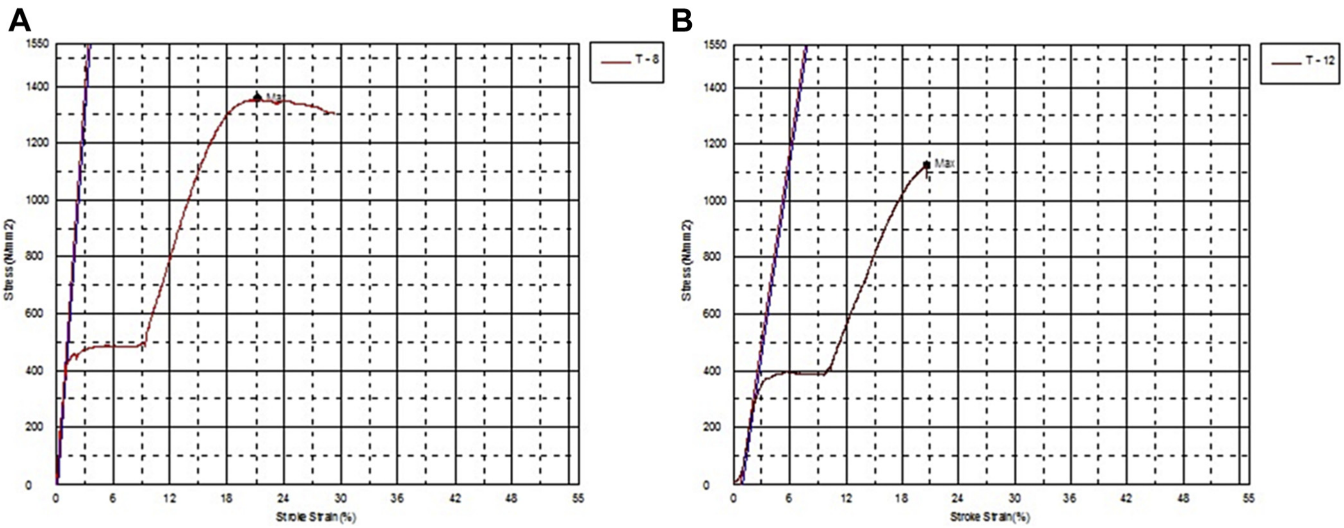

Fig. 2. Stress/strain curves in tensile test at T1. (A) Thermoelastic NiTi wire. (B) Pseudoelastic NiTi wire. The red line in the figures determines the slopes of the plots for continued loading before the thermoelastic region.

working range than TE wires. Similar results were found by Lombardo et al. [6] and Parvizi and Rock [23] in bending tests. Gatto et al. [3] observed that thermally active wires produced significantly lower working forces than superelastic wires of the same size. In addition, Gravina et al. [4] found in a tensile test that heat-activated NiTi archwires presented slighter deactivation loadings than superelastic archwires. In contrast, Quintão et al. [24] found no significant differences in the deactivation forces among PSE and TE wires.

On the other hand, after 3 months of clinical usage (as-retrieved), we observed in the present study that PSE wires continued with significantly higher load at fracture, UTS and YS, maximum tension, and flexural ultimate strength than TE wires. However, the maximum elastic strain (springback) was similar in T1 because the modulus of elasticity decreased in the PSE wires and increased in the TE wires with respect to baseline values. These results suggest that even though TE is still more prone to fracture than PSE wires after clinical usage, both wires' performance and work range are more similar at this time point. The UTS gives the breaking strength to the wire and under normal clinical manipulation and patient use conditions, the stress will not be high enough to cause archwire fracture. However, if the clinicians recycle these wires, they must understand that the decrease of the UTS in as-retrieved wires due to the degradation of the wires exposed to the intraoral environment, make these more vulnerable to fracture. Wire fractures may cause patient discomfort or undesirable tooth movement [25].

Meanwhile, regarding the intragroup differences between as-retrieved and as-received wires (T1-T0), we observed that PSE wires showed a significant decrease in the values of load at fracture and UTS. However, in TE wires, the flexural ultimate strength value increased. These findings suggest that PSE wires are more likely to

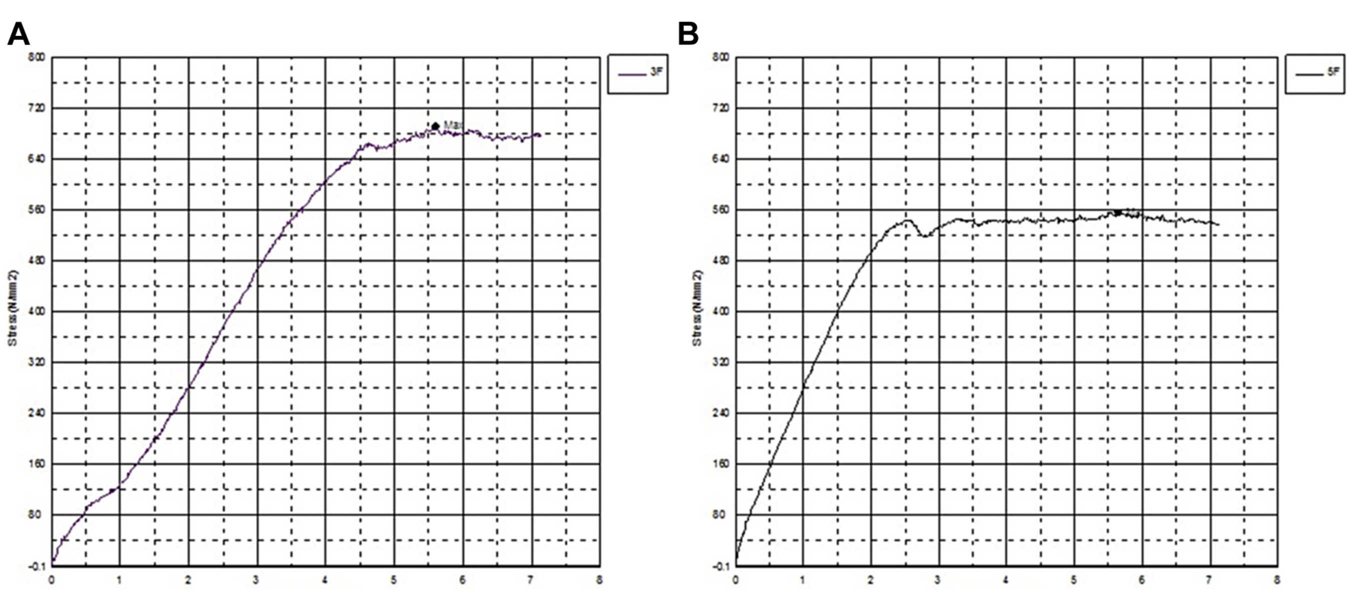

Fig. 3. Stress/strain curves in bending test at T0. (A) Thermoelastic NiTi wire. (B) Pseudoelastic NiTi wire. The red line in the figures determines the slopes of the plots for continued loading beyond the pseudoelastic region.

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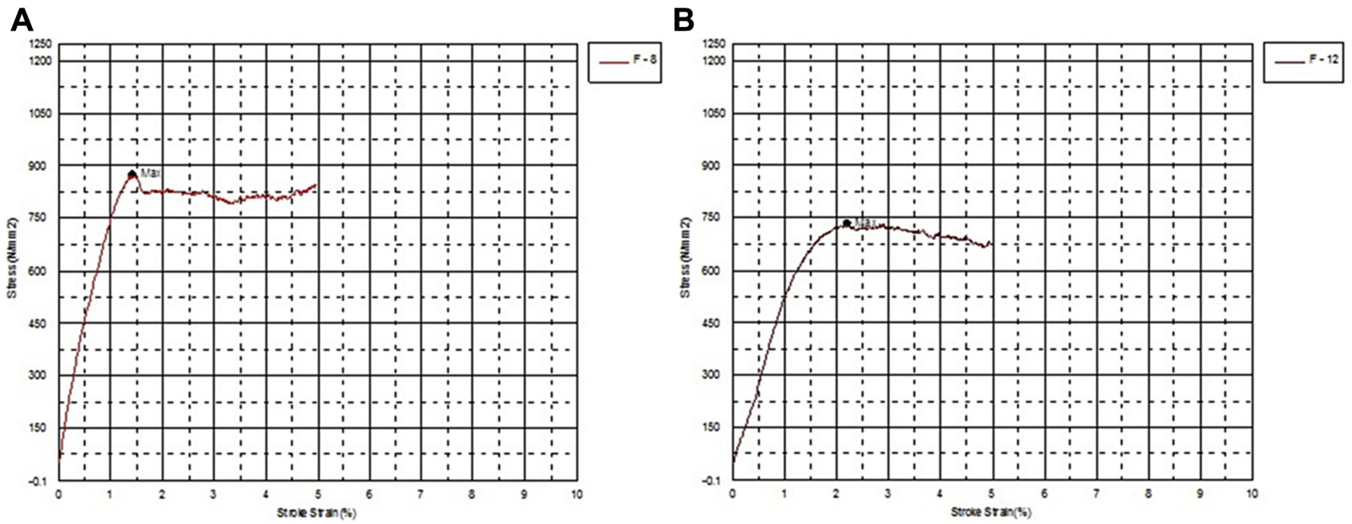

Fig. 4. Stress/strain curves in bending test at T1. (A) Thermoelastic NiTi wire. (B) Pseudoelastic NiTi wire. The red line in the figures determines the slopes of the plots for continued loading before the thermoelastic region.

fracture after clinical usage than as-received wires, whereas TE wires could become stronger than new wires in the oral environment. Likewise, Kapila et al. [26] showed that both clinical recycling and dry heat sterilization produced significant changes in the loading and unloading characteristics of Nitinol and NiTi wires. Bavikati [8] tested the mechanical properties of NiTi wires from three different brands after clinical use and autoclaving and found a significant decrease in the UTS and surface roughness when compared with as-received wires. Pop et al.[16] observed a decrease in the YS, UTS, and elastic modulus in 0.016-inch NiTi wires after 2 months of clinical use, but only YS showed statistically significant differences compared with the as-received wires. In our study, the UTS at PSE wires decreased by 2.37% and that at TE wires increased by 2.92% (16). Likewise, Devaprasad and Chandrasekaran [25] found that SE wires lost 5% of their UTS after 6 months of clinical usage.

Finally, we also found that despite differences in the changes in the mechanical properties after their use in the oral environment of both of the NiTi wires studied, these changes were proportional to each other, and the deterioration that suffered by both alloys showed no significant differences after 3 months of clinical usage.

Regarding clinical implications of our findings, it is important to highlight that TE and PSE NiTi wires are different in their mechanical properties, and it would be wise for clinicians to choose TE wires for patients with more severe crowding or periodontal problems where lower levels of force are needed. Taking into account that their working performance is shorter than that of PSE wires, it may be necessary to replace the archwire earlier. However, the use of PSE wires could be more beneficial for moderate to mild crowding and healthy periodontal patients with a better performance of the archwire during longer time periods and less prone to fracture. It is important to note that clinical studies have not found significant differences in the alignment efficiency of PSE and TE NiTi wires, suggesting that the force on and/or stress developed by the NiTi superelastic archwires under clinical conditions will not be sufficiently high to take advantage of the superelastic property [27–29]. As for the possibility of recycling the SE wires after clinical usage, this is not recommended because of their susceptibility to fracture and loss of properties.

Limitations of the present study are that we tested only one wire size of each type from the same manufacturer and the inability to provide fundamental materials science reasons for the property changes. The selection of 0.014-inch–diameter wires were made for this study considering it to be the most common initial wire in most of the orthodontic techniques.

5. Conclusions

As-received PSE wires were more rigid and delivered higher levels of force than TE wires, but their work range and resistance to fracture were better. After 3 months of clinical usage, both PSE and TE wires lost some of their mechanical properties, such as load at fracture and UTS in PSE wires and flexural ultimate strength on TE wires; however, the changes in the mechanical properties after 3 months of clinical usage were similar for TE and PSE alloys.

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