Evolution of flexural rigidity according to the cross-sectional dimension of a superelastic nickel titanium orthodontic wire

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SUMMARY The choice of the most suitable orthodontic wire for each stage of treatment requires estimation of the forces generated. In theory, the selection of wire sequences should initially utilize a lower flexural rigidity; thus clinicians use smaller round cross-sectional dimension wires to generate lighter forces during the preliminary alignment stage. This assessment is true for conventional alloys, but not necessarily for superelastic nickel titanium (NiTi). In this case, the flexural rigidity dependence on cross-sectional dimension differs from the linear elasticity prediction because of the martensitic transformation process. It decreases with increasing deflection and this phenomenon is accentuated in the unloading process. This behaviour should lead us to consider differently the biomechanical approach to orthodontic treatment.

The present study compared bending in 10 archwires made from NiTi orthodontics alloy of two cross-sectional dimensions. The results were based on microstructural and mechanical investigations. With conventional alloys, the flexural rigidity was constant for each wire and increased largely with the cross-sectional dimension for the same strain. With NiTi alloys, the flexural rigidity is not constant and the influence of size was not as important as it should be. This result can be explained by the non-constant elastic modulus during the martensite transformation process. Thus, in some cases, treatment can begin with full-size (rectangular) wires that nearly fill the bracket slot with a force application deemed to be physiologically desirable for tooth movement and compatible with patient comfort.

Introduction

The last decade has shown that the largest success of shape memory alloys (SMAs) is in the field of bioengineering and biomedical applications. SMAs are materials with an intrinsic ability to remember an initial configuration. Two main unusual behaviours exist: the superelastic effect and the shape memory effect. The first property is the ability to undergo large deformations with zero final permanent strain at a constant high temperature ($>A_f$: finish of reverse transformation of martensite). The second property occurs when material is deformed at a constant low temperature ($<M_f$: finish of martensite transformation) and can be recovered through a heating cycle. The development of nickel titanium (NiTi) in dentistry has contributed significantly to the evolution of orthodontic appliance treatment (Burstone *et al.*, 1985; Miura *et al.*, 1986; Nakano *et al.*, 1999). The most important property used in this field is superelasticity, especially applied in bending. These NiTi wires are able to satisfy the ideal requirements for a fixed archwire appliance, such as high flexibility, minimal plastic deformation, and production of constant light forces over a wide range of displacements (Burstone, 1981; Miura *et al.*, 1986). The values of strength and resilience of these alloys allow a reduction in the number of archwires (Andreasen, 1980).

However, the choice of the most suitable orthodontic archwire for each stage of treatment requires an estimation of the forces generated. In theory, the selection of wire

sequences should allow a progression from lower flexural rigidity in the initial alignment stages of treatment to the greater rigidity required in the final stages for the control of tooth movement. Burstone (1981) calculated the flexural rigidity of a large range of wires at different cross-sectional dimensions to assist clinicians in wire selection. The majority of orthodontists are aware of these assessments and use smaller round cross-section wires and/or incorporate complex loop designs to generate lighter forces during the preliminary alignment stage. These requirements suppose that alloys show a linear relationship between stress (σ) and strain (ϵ), which is described by Hooke's law: $\sigma = E\epsilon$, where the elastic modulus (E) is constant, and does not change with stress and strain. Thus, flexural rigidity (EI, where I is the moment of inertia for a particular cross-sectional dimension) is likewise constant (Brantley *et al.*, 2001). However, these requirements are not always explained explicitly in terms of specific alloy properties. Hooke's law is true for all conventional alloys, but not obligatory for NiTi alloys because of their superelasticity. This refers to a situation where comparatively large strains produced as a result of loading are completely recovered upon unloading at a constant temperature (Miura *et al.*, 1986). NiTi alloys are characterized by a thermoelastic transformation between the austenite and martensite phases. This transformation, which is reversible, is accompanied by a thermal hysteresis, and can be induced by a variation in temperature or stress.

When the formation of stress-induced martensite (SIM) occurs, the result is superelasticity. The most favourable variants of martensite with a shape change in the direction of the external stress outgrow the other variants and are responsible for the large observed deformation. The elastic deformation then becomes non-linear (Guénin, 1986; Brantley *et al.*, 2001).

The aim of this study was to evaluate the flexural rigidity of wire with different cross-sectional dimensions in a loading/unloading cycle, in order to clarify the influence of the effect of superelasticity. A three-point bending test using free-end beam theory was conducted on superelastic NiTi wire test samples. The magnitude of forces and flexural rigidity and their relationship with cross-sectional dimensions are discussed.

Materials and methods

Materials

Ten NiTi orthodontic alloy archwires (NeoSentalloy F200®, Tomy GAC International, Inc., Central Islip, New York, USA), having two cross-sectional dimensions, were selected for the study. The choice of only one manufacturer is appropriate because the mechanical properties of NiTi orthodontic wire products are affected by the chemical composition, process fabrication and heat treatment of the alloy (Guénin, 1986; Jordan *et al.*, 1994; Thayer *et al.*, 1995), and thus will vary among manufacturers. These parameters are not discussed in this study. The samples consisted of five straight wires with a cross-section of 0.018×0.018 inches (0.457×0.457 mm) and five wires with a cross-section of 0.020×0.020 inches (0.508×0.508 mm). The test samples were 30 mm in length. The measurements were carried out using a vernier calliper. The measured cross-sectional dimensions of each sample are summarized in Table 1. The moment of inertia of the cross-section (I) was also calculated, and for a square beam, is given by $I = a^4/12$, where a is the thickness of the square. The moment of inertia value is a geometric quantity that corresponds to the resistance of a particular cross-section to bending (Brantley *et al.*, 2001).

Methods

A three-point bending test was conducted in a manner similar to that previously described (Miura *et al.*, 1986;

Gurgel *et al.*, 2001). The loading and unloading were performed under the same conditions on a testing machine (GT-TesT GmbH Universal testing machine, model 112, Erkrath, Germany) with a span of 14 mm. The mid-portion of the wire segment was deflected to 3 mm at a rate of 2 mm/minute under pressure from a stylus connected to a 20 N load cell. All the measurements were taken in water at a constant temperature of 37°C ($\pm$ 0.5°C) regulated by a thermostatic water bath.

The values of force/deflection with the two cross-sectional dimensions (0.018 and 0.020 inch square) were continuously obtained from the passive position to a total activation of 3 mm and then during deactivation to zero deflection.

Descriptive statistics, including the means and standard deviations (SD), were calculated for each wire tested. Non-parametric tests were used because of the small number of samples ($n = 5$). Indeed, the results of each group were not governed by a normal distribution. For each dimension of archwire, a Wilcoxon paired test was used to assess the differences between loading and unloading. A Mann–Whitney unpaired test was used to compare the two cross-sectional wires during loading and unloading. Significance for all statistical tests was predetermined at $P \leq 0.05$.

The relationship between the applied force (F) and the deflection (δ) under three-point bending conditions is given by Ashby (1992)

$$F = 48 \frac{EI\delta}{L^3} \tag{1}$$

The forces generated in bending are proportional to the deflection, to the modulus of elasticity (E: constant for a conventional alloy), to the moment of inertia of the cross-section (I) and inversely proportional (cube power) to the span length (L) between supports (analogous to interbracket distance). Equation (1) is useful for comparing the relative values of forces in bending for archwires of different alloys and different cross-sectional shapes (Dorlot *et al.*, 1986; Brantley *et al.*, 2001; Gurgel *et al.*, 2001). Consequently, the flexural rigidity for wires with the same span length L is given by EI, where E represents the alloy contribution and I the segment geometry contribution. Because of the dependence of I on the fourth power of thickness for a square cross-section wire, there can be considerable differences in the resistance to bending of wires with different cross-sectional dimensions.

Table 1 Mean (standard deviation), and moment of inertia of archwire test sample groups.

Nominal wire size (mm)	Mean (standard deviation) (mm)	Moment of inertia I (10^{-4} mm^4)	Difference in moment of inertia I (%)
0.457×0.457 (0.018 × 0.018 inch)	0.4495 (± 0.0041)	34.0	↓
0.508×0.508 (0.020 × 0.020 inch)	0.4988 (± 0.0022)	51.6	+51.8

Results

Regarding the orthodontic wires examined, the mean values for the cross-sectional dimensions were approximately 2 SD smaller than their commercial size. The manufacturers intentionally produce smaller cross-sectional dimension wires than the nominal dimensions to facilitate their placement into the brackets (Eliades *et al.*, 2001). Because of the fourth-power dependence on thickness, the moment of inertia increase was approximately 51.8 per cent between the thickness of the 0.018 and 0.020 inch rectangular wires.

The bending curves for each wire are shown in Figure 1. Examination of these two sets of wires shows that the bending curves were characteristic of superelastic behaviour. When a stress is applied above the martensite start temperature (M_S), when referring to the temperature at which the transformation of austenite to martensite begins, a mechanically elastic martensite is stress induced. This phenomenon will disappear when the stress is released (Guénin, 1986; Brantley *et al.*, 2001).

The mean force/deflection values of 0.15, 0.5, 1, 1.5, 2 and 2.5 mm when the total bending was achieved at 3 mm are shown graphically in Figure 2. The mean values represent the forces (in Newtons) necessary to deflect the sample during loading and the forces of these deflections during unloading; these values can be seen in Figure 1.

Two plateau regions are evident. The upper plateau corresponds to the formation of SIM variants. Many variants of SIM are formed and orientated preferentially. The preferred plates are those whose shape strain will permit maximum sample deflection. On unloading, the reverse occurs, and the force/deflection curve follows the lower plateau region. It corresponds to the reverse transformation, in which the martensitic phase is gradually transformed back to the austenitic phase (Wayman and Duerig, 1990).

The slope of the initial and final linear regions corresponds to the rigidity of the austenitic phase in bending. Using equation (1), values of flexural rigidity (*EI*) at loading and unloading deflections of 0.15, 0.5, 1, 1.5, 2 and 2.5 mm can be calculated for the total bending deflection of 3 mm; these values are presented in Figure 3.

By comparison, the modulus of elasticity (*E*) of stainless steel is between 160 and 180 GPa (Brantley, 2001) and the moment of inertia (*I*) for a 0.018 × 0.018 inch archwire is 36.3 × 10^{-4} mm^4 (Brantley *et al.*, 2001). Thus, the corresponding flexural rigidity (*EI*) is between 581 and 653 N mm^2 (Figure 3). This value does not change during bending because there is no change in *E* for a conventional alloy (stainless steel) and *I* remains constant for a given size (Ashby, 1992).

Above 0.5 mm deflection, the stainless steel wire undergoes permanent deformation. However, the flexural rigidity is not constant between loading and unloading for NiTi samples because transformation between the austenite and martensite phases alters the value of *E*.
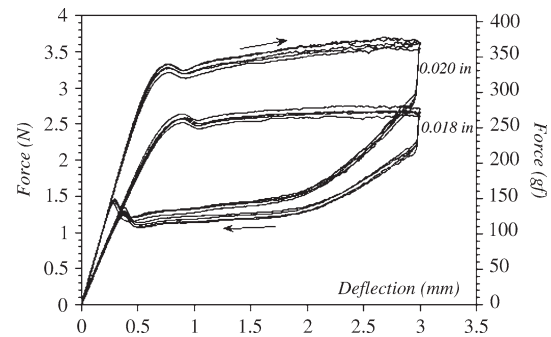

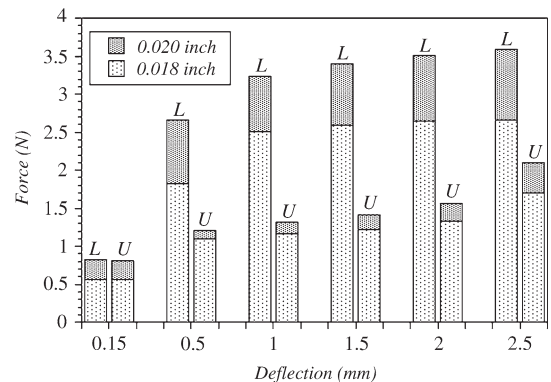**Figure 1** Force/deflection curves for the 0.018 and 0.020 inch wire samples at 37°C.

Figure 2 Mean of forces for the 0.018 and 0.020 inch wires at different deflections (*L* = loaded; *U* = unloaded).
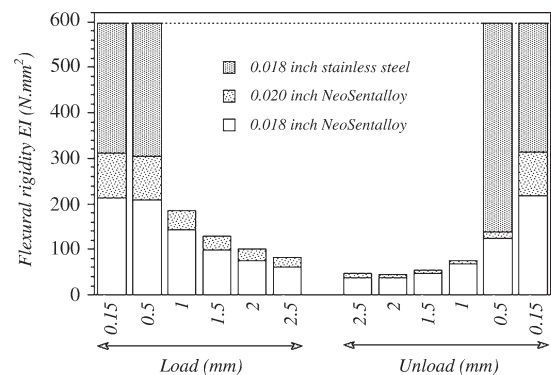

Figure 3 Flexural rigidity as a function of wire size at different deflections.

At 0.15 mm deflection, the behaviour of elasticity was linear for these NiTi wires: the slope of the initial and final regions corresponds to the elastic strain of the austenite phase. The elastic modulus is constant, and does not change with stress or strain. In this region, the level of force (Figures 1 and 2) is the smallest, and the flexural rigidity (Figure 3) for each NiTi wire has its largest value for loading and unloading. For 0.15 mm deflection, the mean force for the 0.020 inch wire was approximately 45.6 per cent greater

than that for the 0.018 inch wire, whereas the moment of inertia increased about 51.8 per cent between these two sizes.

Above a certain level of force, the elasticity behaviour becomes non-linear. The upper plateau corresponds to the formation of martensite with stress (austenite→martensite) and the lower plateau to the reverse transformation (martensite→austenite). The mechanical hysteresis, measured as the difference between the forces of the upper and lower plateaux, increases with cross-sectional dimensions. The values of force for the loading plateau remain approximately constant, but the loading plateau is steeper for the larger 0.020 inch wire size (Van Humbeeck *et al*., 1990; Morgan and Friend, 2001).

Comparing the two cross-sectional dimensions, the 0.020×0.020 inch archwire exhibited a greater load difference between the loading and unloading superelastic plateau than the 0.018×0.018 archwire for 1, 2 or 3 mm deflection. At 1.5 mm deflection, the difference in the force levels on the upper plateau for the two wire sizes was approximately 30 per cent. However, during the unloading process, the difference in forces at the superelastic plateau for the two cross-sectional archwires was clearly less important (≈15 per cent in Figures 1 and 2). Both the 0.018 and 0.020 inch NiTi wires had similar levels of force (between 1 and 1.5 N) during the superelastic unloading process. The mechanical hysteresis, measured as the difference between the forces of the loading and unloading plateaux, increased in the region of 46 per cent from 0.018 to 0.020 inch wire thickness.

When the elastic deformation is non-linear, the flexural rigidity calculated from equation (1) does not remain constant as with a conventional alloy (stainless steel) but decreases significantly during the loading and unloading superelastic plateaux. The important decrease in *EI* is greater for the clinically relevant unloading plateau (Figure 3).

Discussion

For the superelastic NiTi alloy wires used in this study, during loading or unloading, the delivered forces changed slightly when the deflection varied (Figures 1 and 2). At small deformations, the NiTi samples showed linear elasticity similar to conventional alloys: the atoms move very slightly from their initial position, to take up new positions. When the stress is removed, they return to their original positions. The modulus of elasticity is related to the bonding strength between the atoms and is constant. Therefore, *EI* (flexural rigidity) is invariant at the initial loading and final unloading for each of the two NiTi wires (Figure 1). At 0.15 mm deflection, the value of *EI* for the NeoSentalloy® wires was much less (between 215 and 315 N mm^2) than the mean value for stainless steel (581–653 N mm^2). However, the experimental flexural rigidity does not increase with cross-sectional dimension (when *E* is

constant) in the manner expected from equation (1). The experimental values for the moment of inertia are smaller than the theoretical ones because these square wires are rolled from round stock and have rounded corners (Sebanc *et al.*, 1984; Kusy and Stush, 1987; Brantley *et al*., 2001). This can make an important contribution to the value of flexural rigidity.

At sufficiently large deformation, the NeoSentalloy® wires exhibit superelastic behaviour. As previously noted, the superelastic deactivation plateau is lower than the activation plateau and approximately parallel to it for bending deformation. The difference in force levels for the two plateaux is called 'mechanical hysteresis'. The main clinical interest of this hysteresis is that the force delivered to the periodontal structures is much lower than the force necessary to activate the wire.

At small deformations, the alloy is austenitic and microstructurally stable. At an undefined critical force, the martensitic transformation takes place. Thus, the mechanical behaviour of superelastic NiTi alloys is largely under the influence of the martensitic transformation. The upper plateau results from the ability of martensite to accommodate the applied deflection by selecting the most favourably orientated variants along the direction of the strain. Each variant is connected with another variant by a twinning plane, which moves easily upon loading (Van Humbeeck *et al*., 1990; Duerig and Zadno, 1990). Without stress, this martensite transformation is unstable, and samples recover their original shape after unloading. The reverse transformation causes the unloading plateau. As long as the applied stress is not sufficiently great that permanent deformation occurs, the original shape will be recovered completely by the reverse transformation, which involves reverse movement of the interfaces between the austenite and martensite phases. This situation is a crystallographic structural change, in which the superelastic deformation results from phase transformation between austenite and martensite and is not a stretching of bonds that occurs during elastic deformation. The growth of the most favourable martensitic variants accommodates the applied stress. This phenomenon requires much lower energy than Hookean elasticity in this temperature and stress range.

During martensitic transformation (forward and reverse), the elastic modulus (*E*) or flexural rigidity (*EI*) is not constant and decreases greatly (Figure 3). The considerable decrease is more important for the unloading superelastic process, as tooth movement occurs during deactivation of the archwire. Also, it is not the relative concentration of the two phases in the NiTi alloy that will determine the resultant rigidity (modulus of elasticity, *E*) of the sample but the process of martensitic transformation.

The response of a NiTi archwire is controlled not only by chemical-free energy but is also dependent on micro-structural factors. Elastic energy is stored in the alloy through a non-dissipative process during forward transformation. This energy is reversible, and corresponds

to elastic strain energy and the interface energies (between austenite and martensite; and between variants of martensite). However, some energy is dissipated by friction due to the movement of interfaces and the formation of defects. These different energies are the controlling factors in the profile of the force/deflection curves in the superelastic range (Otsuka and Ren, 1999).

The loading plateau

The slope of the upper plateau is a function of wire size. The slope is steeper with increasing cross-sectional dimensions (Figure 1). This phenomenon corresponds to increasing elastic strain energy, i.e. the volume fraction of transformed martensitic and frictional energy induced by the transformation are higher. Therefore, the applied stress required to transform austenite to martensite gradually increases with the volume fraction of transformed martensite (Van Humbeeck *et al*., 1990; Morgan and Friend, 2001).

The unloading plateau

The unloading process is controlled by the stored energy, which assists the reverse transformation (Duerig and Zadno, 1990; Tamura and Wayman, 1992; Otsuka and Ren, 1999). The positions of the variants in the deformed state are not stable without stress, and thus they return to their original positions during unloading (Duerig and Zadno, 1990). Consequently, the force level for the unloading plateau is considerably less than that for the loading plateau (mechanical hysteresis). This phenomenon is amplified with higher cross-sectional dimensions because the volume fraction of transformed martensite is more important (Otsuka and Ren, 1999). Thus, the superelastic unloading force levels for each wire (0.018 and 0.020 inch) are very similar, although a large difference exists between the moment of inertia (Table 1).

Conclusions

In this investigation, two different cross-sectional dimensions of a superelastic NiTi archwire showed a smaller force difference for the same deflection during the unloading process, compared with the loading process. This result shows that the moment of inertia, and thus the cross-sectional dimensions of these archwires, is not a factor of primary importance for the level of force. The effect of the moment of inertia exists, but its significance is reduced by the large decrease in the modulus of elasticity (E) arising from martensitic transformation (for the two superelastic plateaux). The flexural rigidity is not strongly linked with the cross-sectional size when there is superelastic behaviour during unloading. Consequently, higher cross-sectional dimension wires that exhibit martensitic transformation allow the required biomechanical objectives of orthodontic treatment.

Simultaneous alignment, levelling, and torque at the beginning of treatment, which are the treatment goals, can be achieved with satisfactory adaptation of the wire in the bracket slot, which will enable the achievement of tooth movement (Gurgel *et al*., 2001).

The estimation of forces produced by superelastic NiTi should be based on an understanding of the physics of the transformation process and its relationship to the value of flexural rigidity of the archwire. In this way, this latter biomechanical parameter could be controlled with improved efficiency.

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