REVIEW ARTICLE

Pseudoelasticity and thermoelasticity of nickel-titanium alloys: A clinically oriented review. Part II: Deactivation forces

Margherita Santoro, DDS, MA,[a] Olivier F. Nicolay, DDS, MS,[b] and Thomas J. Cangialosi, DDS[c]
New York, NY

The purpose of this review was to organize a systematic reference to help orthodontists evaluate commonly used orthodontic nickel-titanium alloys. Part I of the article reviewed the data available in the literature regarding the temperature transitional ranges of the alloys. The thermomechanical behavior of these compounds is, in fact, strictly dependent upon the correlation between the temperature transitional range and the oral temperature range. Part II focuses on the mechanical characteristics of the alloys, such as the magnitude of the forces delivered and correlations with the temperature transitional range and oral temperature. (Am J Orthod Dentofacial Orthop 2001;119:594-603)

The first part of this review analyzed the transitional ranges of commonly used orthodontic alloys.[1] This, the second part, will address the amount of force delivered by the wires to the dentoalveolar structures in the deactivation phase.

The advantage of using light continuous forces during treatment has been demonstrated through animal experiments and histological studies.[2-8] Light forces limit bone hyalinization and reduce undermining resorption, thus allowing for a more physiological tooth movement. To facilitate control of deactivation forces, new materials, including nickel-titanium (NiTi) alloys, have been developed. It is possible to modify the stiffness of these archwires by varying the atomic composition instead of the diameter.[9,10]

A review of the literature about the delivery forces of superelastic wires, however, reveals a large variation in study designs. The methods used to measure force, for example, are not uniform. Forces may be given in centinewtons or newtons or expressed as strengths and given in pascals, whereas in clinical practice we are accustomed to expressing force in grams or ounces.

From the Division of Orthodontics at the Columbia University School of Dental and Oral Surgery, New York, NY.
[a]Assistant Professor of Clinical Orthodontics.
[b]Associate Professor of Clinical Orthodontics.
[c]Professor and Chairman.
Reprint requests to: Margherita Santoro, Division of Orthodontics, Columbia University, School of Dental and Oral Surgery, 635 W 168th St, P O Box 20, New York, NY 10032; e-mail, ms190@columbia.edu.
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Moreover, the parameter commonly used to study and describe force delivery is the stiffness value, that is, the force required to obtain a deflection in the wire below the yield point. At oral temperature, a stainless steel braided wire presents a stiffness of 0.06 (the reference stiffness of stainless steel is equal to 1), whereas NiTi wires, during large deflections, have a stiffness of 0.07.[11] This type of stiffness value does not provide immediately useful information about the amount of force actually released to the dentoalveolar structures.

Force delivery is also influenced by other factors such as alloy composition, cross section, and the number of strands of the wires used in the experiments. Therefore, specific experiments aimed at obtaining data of direct clinical interest need to be designed.

DEFLECTION SETTINGS

In early experiments, uniaxial tensile tests were considered to be the most acceptable method for comparing the mechanical properties of different alloys. In tensile tests the wire is usually attached at one end and stretched. A stretch of more than 8% may be required to see the expression of the superelastic properties in NiTi wires.[11-13] Tensile forces, however, are not usually observed in clinical orthodontic applications. Furthermore, tensile tests generally produce values of the modulus of elasticity, which are significantly lower than those obtained through deflection tests, such as cantilever bending tests.[13]

In flexural tests using a cantilever configuration, delivery forces are evaluated as bending moments and are expressed in grams per millimeters. The deflection generated is measured in degrees.[11] The cantilever-type test is, at present, the standard, ADA-approved method of testing the mechanical properties of alloy, according to ADA specification number 32. It must be taken into consideration, however, that a single direction deformation induced by unilateral bending tests (including tensile, 1-point, or even 3-point cantilever bending models) may simulate superelastic behavior even if the alloy does not possess superelastic properties.[12] Unilateral bending models are, in fact, based on the free-end beam principle, according to which, the wire is allowed to slide unrestrained over the supporting devices. The friction generated when the wire slides over the supports increases the loading force so that superelastic behavior can be simulated even in nonsuperelastic wires, especially if the wire presents a high coefficient of friction, as do titanium-molydenum alloys.[14,15] Also, nonrestrained tests do not take into consideration the loading effect of the friction generated by the ligatures and are therefore unable to properly reproduce the clinical constraint of the wire in the bracket slot.

To obtain reliable and valuable data, researchers must design experimental models that simulate as closely as possible the orthodontic intraoral clinical setting. The method of ligation of the wire to the brackets should be consistent. The interbracket distance, type of bracket used, and length of the wire specimen should be consistent as well. The results obtained in a laboratory should ultimately be compared with the results of analogous clinical trials.[16]

The 3-bracket bending test is a partially restrained model that should provide immediately useful results for orthodontic clinical needs. This model has been used in some recent laboratory experiments performed on austenitic superelastic wires. A common finding of those studies is a relatively high stiffness obtained if the wire is deflected less than 2 mm.[10,14,15,17] Because of the lack of formation of stress-induced martensite (SIM), austenitic NiTi presents a stiffness of 0.28 for small deflections, a value surprisingly higher than the 0.20 stiffness of a classic work-hardened alloy like Nitinol.[11] If the threshold of 2 mm of minimum activation has to be considered reliable, a superior performance of superelastic austenitic NiTi during the alignment phase of treatment seems to be limited to cases of severe dental crowding. This peculiar behavior of austenitic superelastic wires can be more easily explained through the analysis of stress-strain curves (Fig 1).

When the austenite is transformed into SIM, a horizontal plateau appears as an indicator of the expression of superelastic or, more properly, *pseudoelastic* properties.[18-21] In general, a consistent presence of martensite in the alloy, either thermally generated or stress-generated, is responsible for the lowering of the delivery force. Martensite is, in fact, the low stiffness phase and presents a modulus of elasticity of 2,000,000 psi; austenite is the higher stiffness phase with an elastic modulus of 8,000,000 psi. The relative concentration of the 2 phases in the alloy will determine the resultant stiffness of the wire and the amount of force delivered. The modulus of elasticity of a work-hardened martensitic wire (original Nitinol; 3M Unitek, Monrovia, Calif) is 4,800,000 psi, and can be used as a comparison.[22,23]

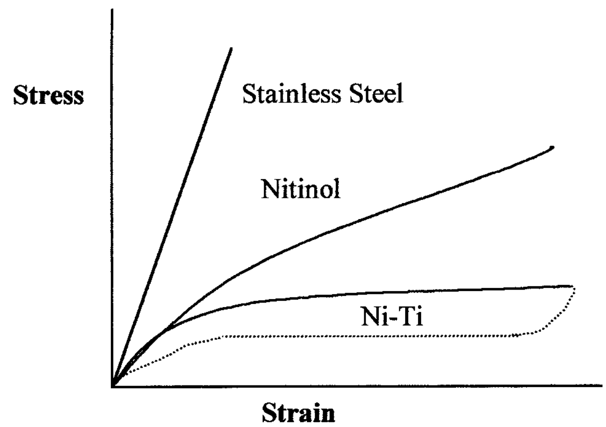

Fig 1. Stress-strain curves for stainless steel, Nitinol, and NiTi.

Superelastic compounds generally present a high stiffness in the initial segment of the slope of the stress-strain graph when the deflection of the wire is still minimum. The initial activation force required for austenitic NiTi can be 3 times greater than the force required to deflect a classic work-hardened martensitic wire. However, once the SIM is formed, the horizontal plateau appears and the alloy "absorbs" any additional load stress and releases it in constant amounts during the deactivation phase. Actually, the linear region corresponding to the deactivation plateau is lower than the activation plateau and parallel to it. This phenomenon is called *hysteresis.* The main clinical interest of hysteresis is that the force delivered to the periodontal structures is lower than the force necessary to activate the wire.

Some confusion about these relatively simple molecular mechanisms arises from the erratic results obtained on the behavior of the alloys when different experimental setting designs are used. For example, Nakano et al[15] performed a 3-point bending test at 37°C on several available superelastic and work-hardened alloys with a maximum bending deformation of 2 mm. The results can be compared to those obtained with a 3-bracket bending test performed at 35°C on the same alloys by Segner and Ibe[14] and Oltjen et al.[10]

A .016 × .022-in Nitinol SE (3M Unitek) wire, an alloy deprived of pseudoelastic properties according to

Table I. Force delivery characteristics of some of the commercially available NiTi alloys*

Product	*Manufacturer*	*Force delivery (loading/unloading)†*		*Experimental setting*	*Reference*
Japanese NiTi	Furukawa Electric	Heavy .016 280 g Medium .016 170 g Light .016 70 g		3-point bending test 2-mm deflection	12
Thermomemoria	Leone	.016 .018 .018 × .025 .019 × .025	312/60 g/mm 452/87 g/mm 916/130 g/mm 1018/1140 g/mm	3-point bending test 1.6-mm deflection	34, 35
Nitinol SE	3M Unitek	.016 × .022 300 g		3-point bending test 2-mm deflection Unloading 37°C	15
		.016 × .022 203 g/mm		3-bracket bending test Unloading 2-mm deflection 35°C	14
Nitinol Classic	3M Unitek	.017 × .025 79.6 g/mm		3-bracket bending test 2-mm deflection Room temperature	10
		.016 × .022 307 g		3-point bending test 2-mm deflection Unloading 37°C	15
		.016 × .022 215 g/mm		3-bracket bending test Unloading 2-mm deflection 35°C	14
		.016 790 g		3-point bending test 2-mm deflection	12
27°C Superelastic Copper Ni-Ti	Ormco	.016 × .022 137 g		3-point bending test Unloading 2-mm deflection	15
35°C Thermo-Active Copper Ni-Ti	Ormco	.016 × .022 80 g		3-point bending test Unloading 2-mm deflection 37°C	15
40°C Thermo-Active Copper Ni-Ti	Ormco	.016 × .022 87 g		3-point bending test Unloading 2-mm deflection 37°C	15

*1 g = 0.98 cN.
† Wire sizes are measured in inches.

Segner and Ibe,[14] releases 300 g/2 mm of force in a 3-point bending test,[15] but delivers 203 g/mm in a 3-bracket bending test. According to other authors, the same Nitinol SE should release a much lower force of 79.6 g/mm at room temperature; this difference in the amount of force delivery cannot be explained simply by an increase, at room temperature, of thermally induced martensite due to the phase transition.[10]

To increase confusion about the reliability of experimental results, some variability in force delivery has also been noticed in work-hardened alloys like the classic Nitinol. A work-hardened alloy should, theoretically, be insensitive to modifications of experimental design because it should not undergo any noticeable phase transformation. However, Nitinol delivered a higher force (215 g/mm) in the 3-bracket bending test versus the 3-point bending test (307 g).[14,15]

The inconsistency of the results available for Nitinol limits the reliability of the conclusions regarding superelastic alloys. This inconsistency has, in the past,

Table I. Cont'd

Product	*Manufacturer*	*Force delivery (loading/unloading)***	*Experimental setting*	*Reference*
Ni-Ti	Ormco	.016 × .022 313 g	3-point bending test Unloading 2-mm deflection 37°C	15
		.016 102 cN at 35°C 112 cN at 50°C	3-point bending test Unloading 3-mm deflection	43
Sentalloy	GAC International	Heavy .016 × .022 293 g Medium .016 × .022 193 g Light .016 73 g	3-point bending test Unloading 2-mm deflection 37°C	15
		Light .016 62 cN at 50°C 20 cN at 35°C Heavy .016 108 cN at 50°C 77 cN at 35°C	3-point bending test Unloading 3-mm deflection	43
		Heavy .016 × .022 879 g/mm Medium .016 × .022 695 g/mm Light 203 g/mm	3-bracket bending test 2-mm deflection 35°C	14
Neo Sentalloy	GAC International	F240 .016 × .022 143 g F80 120 g	3-point bending test 2-mm deflection Unloading 37°C	15
		F100 180 g/mm F200 250 g/mm F300 250 g/mm	37°C Unloading	37
		F240 .016 × .022 36 g/mm F80 48 g/mm F160 42 g/mm	3-bracket bending test 2-mm deflection Unloading 35°C	14
Heat Activated NiTi	Highland Metals	g/mm loading 20°C = 432 36°C = 480		34, 35
Reflex Heat Activated	TP Orthodontics	.016 × .022 159.5 g/mm .018 × .025 178 g/mm	1-mm deflection	34, 35

led to a certain skepticism regarding the advantage of using NiTi during the alignment phase when compared, for example, with well-established multistranded stainless steel wires.[10,14,24-34]

TEMPERATURE SETTINGS

Force delivery of thermoelastic compounds such as austenitic NiTi and copper NiTi is strictly dependent on the temperature. The control of the temperature in the experimental setting requires proper equipment such as thermostats and insulated chambers.[35] As a simpler alternative approach, numerous experiments have been performed at room temperature. However, one has to consider that wires with a temperature transitional range (TTR) located at average oral temperature (35°C) are partially martensitic, so that in mechanical tests conducted at room temperature they will perform as superelastic wires and deliver low forces. At a higher oral temperature, because of the tendency of the grain structure to be reconverted from martensite to austenite, a sufficient formation of martensite and SIM may be prevented in these wires.[36]

The highest temperature at which SIM can form is conventionally defined as *Md*. In active austenitic orthodontic alloys, Md is usually greater than the austenite final, or Af, temperature; Af is located slightly below the oral temperature.[37] The so-called martensitic active wires have a TTR located at oral temperature, so that a greater amount of martensite is constantly available. This type of alloy presents both thermoelastic and pseudoelastic properties and can be considered the long-awaited NiTi alloy hypothesized by Andreasen[38,39] and Otsubo.[40]

A general explanatory example of the combined thermal and mechanical behavior of austenitic orthodontic alloys is provided by 3-point bending tests performed at nominal mouth temperature (35°C). With the use of this test, Tonner and Waters[41] found an increase of the plateau of the delivery forces with the increasing temperature. For Sentalloy light .016-in wire (GAC International, Islandia, NY), the deactivation force increased with the temperature from 20 cN at 35°C to 62 cN at 50°C. For Sentalloy heavy .016-in wire (GAC), the deactivation force increased from 77 cN at 35°C to 108 cN at 50°C. For .016 Ormco Ni-Ti (Ormco, Orange, Calif), the deactivation plateau increased from 102 cN at 35°C to 112 cN at 50°C. Although the results might not be directly applicable to clinical situations (a 3-bracket test would have been more appropriate), the authors confirmed the decrease of SIM with the increase in temperature. They also confirmed that superelastic alloy wires require a deflection of at least 2 mm over a span of 13 mm before superelastic behavior is detected. Austenitic alloys are therefore superelastic mainly when used for the correction of gross misalignments of teeth.[41,42]

STRESS-STRAIN VALUES: HOW HEAVY ARE THE DEACTIVATION FORCES?

An orthodontic force can be defined as heavy or light according to the ratio between the magnitude of the force applied and the affected root–bone surface.

Because of the surface areas of the periodontal structure involved, an ideal archwire should be able to deliver differential forces to the arch segments. The force should range from about 70 g to 80 g in the incisor area and gradually increase toward the posterior segments, up to 300 g.[2-8,43,44]

Forces below the threshold of 100 g are, in general, released by very small diameter multistranded stainless steel wires and by superelastic wires under specific conditions of phase transition from austenite to martensite. However, not all the experimental settings are able to trigger proper superelastic behavior in superelastic wires. Therefore, one of the objectives of this review was to identify properly designed, published studies that provide data with immediate clinical usefulness.

Unloading force values available from previous studies, with a brief description of the experimental settings, are reported in Table I.

Because of difficulties in manufacturing procedures, great variability of elastic properties is found among austenitic wires provided by different distributors and even among wires from different batches from the same distributor. Some authors have suggested that the distributors should at least quantify the elastic parameters of each batch if they are not able to improve the standardization of the mechanical properties of the wires.[44]

As mentioned before, sometimes the variability in the data can be due to the different experimental designs used in the studies. For instance, numerous data are available on Sentalloy, one of the most popular alloys. In a 3-point bending test at 35°C, a .016 × .022-in Sentalloy medium wire delivered 193 g/mm, whereas in a 3-bracket bending test, the same wire released 695 g/mm. The discrepancy between the data is evident. If, for particular reasons, the clinician must use this type of austenitic alloy with a low TTR, then, to maintain the level of delivery forces below 100 g, it will be necessary to select a wire with a smaller diameter, such as .016-in round wire. Such a wire delivers 73 g of force in a 3-point test[15] or, according to other studies,[41,42] 20 cN (equivalent to 20.6 g/mm).

For Neo Sentalloy (GAC), the data available on forces produced are more consistent with clinical orthodontic situations. In a 3-point bending test at 37°C, a .016 × .022-in wire, type F240, delivers a force of 143 g/mm.[15] In a 3-bracket test, the same wire delivers a much lower and more physiologic force of 36 g/mm, as long as at least 2 mm of deflection are applied.[14]

As an alternative to austenitic NiTi, ternary alloys like Copper Ni-Ti (Ormco) have been recently proposed. Rectangular Copper Ni-Ti wires, .016 × .022 inches in diameter and superelastic at 35°C and 40°C, seem to consistently produce light forces in the range of or below 100 g/mm when tested in a 3-bracket bending system at oral temperature (37°C). Because of the more distinct temperature-related reconversion of SIM in austenite, 27°C Superelastic Copper Ni-Ti (Ormco) releases a higher force of 137 g/mm.[14] Therefore, for 27°C Superelastic Copper Ni-Ti, as for Sentalloy, it will be necessary to select a small diameter, preferably round, wire.

MULTISTRAND STAINLESS STEEL VERSUS NICKEL-TITANIUM

When comparing stainless steel and nickel-titanium of similar size or of similar number of strands in laboratory tests (within-category comparisons), the results show, in general, considerably greater stiffness for stainless steel—about 4 to 5 times higher than that of superelastic NiTi.[10,30,31]

Oltjen et al[10] analyzed several NiTi alloys using a 3-bracket bending test at room temperature, deflecting the wires by up to 3 mm. With 2 mm of deflection, an austenitic NiTi wire (.017 × .025 in) delivered only 18.0 g/mm, whereas a multibraided 9-strand NiTi wire (.017 × .025 in) delivered 12.4 g/mm. However, it must be considered that the low temperature at which the experiment was performed favors the formation of martensite and lowers the forces delivered by NiTi wires. At oral temperature, the same wires would deliver higher forces, perhaps forces close to those delivered by alloys other than NiTi. Actually, a classic multistrand stainless steel wire, a .017 × .025-in 8-strand, delivers 170.7 g/mm of force, and a .0175-in 6-strand stainless steel wire delivers 43.1 g/mm. These forces are not temperature dependent and are considered within the physiological range. With the increase of deflection to 3 mm, a .0175-in 3-strand stainless steel wire delivers 160.3 g/mm of force, which is a much higher but, perhaps, a reasonable value, if we consider that a .016-in Chinese NiTi (austenitic alloy) wire delivers 199.3 g/mm of force.[10]

The difference in performance between a NiTi alloy and a multistranded stainless steel is detectable mainly with the increase of loading. For example, at 2 mm of deflection, the stiffness of a .0175-in 6-strand stainless steel wire (43.1 g/mm) is more than double the stiffness of a .017 × .025-in NiTi wire (18.8 g/mm).[10]

Aside from the acceptable delivery forces, the remarkable clinical advantage of superelastic wires is the exceptional springback and the resistance to permanent deformation over a long period of time, characteristics that are not observed in stainless steel or even Nitinol.[11] Multistranded stainless steel, Nitinol, and Chinese NiTi have been compared after extended use during the alignment stage to test the degree of permanent deformation.[16] More than 50% of the braided wires presented a high degree of permanent deformation, to the point that the wire was not reusable, whereas neither Nitinol nor Chinese NiTi showed any deformation. The results of the clinical study were confirmed with the use of a 3-point bend test performed at room temperature, where the wires were subject to a deflection up to 4 mm. At 2 mm of deflection, the braided wire already showed a moderate plastic deformation, which means that the springback of stainless steel was not sufficient to guarantee optimal performance over a long period of time.

According to these data, under conditions of minimum crowding there is no special reason to use a superelastic alloy wire rather than an established multistranded stainless steel wire, because the range of force delivered by the multistranded stainless steel is considered acceptable. Superelastic NiTi may represent the elective choice when moderate crowding is present and when arch form and torque control are required in the initial stages of treatment because an equivalent rectangular multistranded stainless steel wire presents rather high stiffness and is subject to permanent deformation. In cases of severe dental crowding and in periodontally compromised patients, when the amount of force delivery is a concern along with the need for torque control, a rectangular Ormco Copper Ni-Ti 35°C and 40°C is indicated even more than austenitic superelastic NiTi.

GRADED THERMODYNAMIC NITI

In an early experiment, Miura et al[12] immersed several specimens of superelastic NiTi in a nitrate salt bath at different temperatures for increasing heating intervals and then tested their mechanical properties with a bending test. A heat treatment at 500°C for 20 minutes decreased force delivery in Japanese NiTi to a plateau of 50 g, compared to a plateau of 300 g in untreated wires. A higher temperature treatment generates a complete loss of the superelastic properties.[12]

A few years later, direct electric resistance heat treatment (DERHT) of the NiTi wires was proposed. The heat treatment of selected sections of the archwire by means of different electric currents delivered by electric pliers modified the values of the deactivation forces by varying the amount of austenite present in the alloy.[45-47] After heating the anterior segment for 60 minutes, the linear plateau of the deactivation force dropped to 80 g in a 3-point bending test at room temperature. Similar manufacturing procedures have been perfected to produce wires such as Bioforce Sentalloy (GAC) that are able to deliver selective forces according to the needs of the individual dental arch segments.[47] Evans and Durning[18,19] classified wires like Bioforce as phase V or graded thermodynamic NiTi.

Three varieties of superelastic alloys available from GAC (Neo Sentalloy F100 and F200) have been analyzed at 5°C and 37°C in a 3-point bending test and compared to Bioforce. The values of the deactivation plateau roughly corresponded to the official values provided by GAC (100 g/mm in the anterior region, 300 g/mm in the posterior regions of the archwire). Bioforce showed delivery force values close to Neo Sentalloy F100.[37]

TORSIONAL BEHAVIOR

The use of rectangular NiTi wires has been proposed to obtain early torque control during the aligning phase of treatment. However, in terms of sensitivity to interbracket distance, torsional stress is considerably different from flexural stress. In flexural stress, modifications in the length of the specimen generate changes in the forces delivered according to the third power law. In torsional stress, length modification generates linear changes in the forces. As a consequence, the results available from experiments studying flexural behaviors cannot be indiscriminately applied to the torsional behavior of NiTi wires.[48-52]

Filleul and Jordan[48] used differential scanning calorimetry, a technique that measures the enthalpy of a phase transition, to study the torsional behavior of .017 × .025-in sections of Neo Sentalloy F100, and 35°C and 40°C Thermo-Active Copper Ni-Ti (Ormco). The measurements were taken at 22°C, 39°C, and 44°C. In Neo Sentalloy F100 at 22°C, the superelastic plateau appeared with the application of 910 g/mm of torque, whereas at higher temperatures (33°C and 44°C), the alloy was always austenitic and the plateau did not appear. Copper Ni-Ti 35°C and 40°C showed almost equivalent torsional behaviors. At 21.7°C, both presented a superelastic plateau at 560 g/mm; at 39°C the plateau rose to 1190 g/mm. At 44°C, even in the presence of the maximum torque allowed in the experiment, (1400 g/mm), no significant formation of martensite was observed in either alloy.[48] In comparison, a stainless steel rectangular .019 × .025-in wire, for 30° of constant twist, delivers about 3000 g/mm, independent of temperature variations.[52,53] Clearly, the deactivation force released by superelastic NiTi is definitely lower than that released by an equivalent rectangular stainless steel wire, but the property is due more to the intrinsic elastic properties of NiTi compounds than to the presence of a phase transformation.

It should be possible, however, to reduce NiTi torquing forces by lowering the temperature. It is actually a common clinical practice to prescribe cold rinses to increase the amount of martensite present and relieve the pressure exerted by the archwire on the periodontal structures. Meling and Odegaard[49] analyzed the thermodynamic torsional properties of superelastic NiTi during short-term (up to 10-second) temperature changes. The torque apparatus simulated the stress generated by a .018-in slot edgewise bracket, with an interbracket distance of 4 mm. An increase in the temperature generally increased the torsional force delivered, thus confirming a greater presence of the stiffer austenitic phase. Cooling procedures, however, reduced the torque level by up to 85% of the baseline. The force variations were generally transient, even if, in some cases, more than 60 minutes were necessary to reestablish the baseline torque. The baseline torque level for 20° of constant twist at 37°C was located at about 1000 g/mm for 27°C Superelastic Copper Ni-Ti, 35°C and 40°C Thermo-Active Copper Ni-Ti, Neo Sentalloy, and Rematitan (Dentaurum, Ispringen, Germany). An increase in the delivery force of up to 1500 g/mm, proportional to the increase in temperature, was more obvious in Neo Sentalloy and Rematitan than in Copper Ni-Ti, which remained almost stable. On the other hand, a lowering of the temperature generated a significant decrease in torque level, especially in 40°C Thermo-Active Copper Ni-Ti, in which the torque level markedly dropped to 200 g/mm. Unitek Nitinol SE presented a torque baseline of less than 1500 g/mm, with some increase on heating and a nonsignificant decrease on cooling. Titanol (Forestadent, Pforzheim, Germany) and Elastinol (Masel Enterprises, Bristol, Pa) presented a higher torque baseline, around 200 g/mm, with peaks of 2500 g/mm on heating and not less than 1500 g/mm on cooling.[49]

CLINICAL TRIALS

Randomized clinical trials are necessary to test the hypothesis of faster tooth alignment and minimum discomfort for patients using NiTi wires compared with other alloys. Only a statistically significant improvement in the clinical performance of a new (and more expensive) material can justify the replacement of an established material like stainless steel.[33]

According to experimental data obtained on rats by Warita et al,[30] the rate of tooth movement using light continuous forces was almost double that observed with heavier dissipating forces. Histologically, normal osteoblasts and osteoclasts were observed in the periodontal ligament of rats when superelastic NiTi wires were used, whereas hyalinization and a decrease in the number of cells present was a common finding in the group with a work-hardened alloy.[34] Even considering that the forces per unit area applied to the periodontal structures of rats are quite different from those applied to the human periodontium (because of obvious root dimension differences), the results of the investigation justify more extensive use of superelastic wires.

West et al[29] compared the performance of .015-in multistranded stainless steel wire with .014-in austenitic NiTi wire during a 6-week clinical trial. Only in the lower incisor segment did the superelastic wires show a better performance due to a reduced bracket distance and the formation of SIM. No threshold of crowding was found in which one arch performed better than the other, and no difference was found between the performances of the materials in any other section of the dental arch. The authors concluded that even the statistically significant difference found for the lower anterior segments translated into only a marginal clinical improvement of the alignment.

Similar conclusions were reached by O'Brien et al[28] in comparing austenitic superelastic wires and work-hardened alloys. In a pilot study with a 3-point bending test, the .016-in austenitic wire required more than 600 g/mm of bending moment for the appearance of a superelastic plateau. Tooth movement during the alignment phase was then evaluated over a period of 35 days through a computerized analysis of the distances between the anterior contact points and reference points located at the palatal rugae. No difference in the rate of tooth movement was detected between the austenitic NiTi and the work-hardened alloy.[32]

The performance of austenitic NiTi does not seem outstanding even when a very small diameter is selected for the wires. Jones and Richmond[25] performed an analogous test to compare a triple-strand .015-in stainless steel wire with a .014-in Sentalloy wire and reached the conclusion that the difference in the performance of the wires was clinically undetectable.

Another proclaimed clinical advantage of NiTi wires is the minimum discomfort experienced by the patient during treatment. This assumption is based on the classic histological studies on tooth movement by Reitan and Storey,[2,7] from which it was deduced that there is a definite relationship between the amount of force applied and the pain experienced by the patient.

Some authors, however, have expressed serious doubts about this theory.[3,27] Pain is a multifactorial phenomenon involving several individual psychological and emotional variables and is not readily evaluated by simple statistical evaluation.[25] In a randomized clinical trial, Jones and Chan[24] recorded the pain and discomfort related to orthodontic treatment on a daily basis, using a 100-mm visual analogue scale at 4 periods every 24 hours. Pain after tooth extraction was used as a measure of comparison to grade discomfort related to orthodontics. The 2 wires compared were a .014-in heavy Sentalloy and a multistranded stainless steel .015-in Twistflex (3M Unitek). Surprisingly, the pain reported by the patients was more intense and of longer duration after the insertion of the archwires than after the dental extraction. However, the experience of pain showed a consistent individual threshold; the more pain was experienced after extraction, the more pain was experienced after the insertion of archwires. No difference was found between the archwires or between a maxillary or mandibular archwire. The initial degree of crowding did not affect the degree of pain. Instead, a significant correlation was found between age and the level of pain, with pain increasing with the age of the patient.

It must be considered that the studies mentioned before have been performed with austenitic NiTi with a low TTR, such as Sentalloy. More studies are needed to evaluate the performance of thermodynamic alloys, such as Copper Ni-Ti, or more recent austenitic alloys with a higher TTR, such as Neo Sentalloy.

SUMMARY AND CONCLUSIONS

The rationale for making an educated clinical choice of a NiTi alloy includes two primary considerations:

1. An appropriate stress-related TTR, corresponding to the oral temperature range; this characteristic was discussed in the first part of this review.
2. A physiological level of force delivered to the teeth and surrounding structures; the delivery force is strictly correlated to the presence of martensite in the alloy and is therefore dependent on the TTR as well as on the amount of stress induced.

The experiments designed to study NiTi force delivery levels should take into consideration:

1. The type of loading applied to the wires; it should reproduce as closely as possible a clinical situation. Three-bracket bending models represent the most consistent design.
2. The amount of loading used to test the superelastic behavior; at least 2 mm of deflection are necessary for the formation of SIM in austenitic wires. A deflection below the 2-mm threshold may translate into a higher force delivery correlated with the constant presence of the stiffer austenitic phase. An optimal performance of austenitic superelastic NiTi wires will be obtained in cases of severe dental crowding, when an accentuated deflection due to the irregular interbracket span will generate SIM in a localized area of the arch, usually the lower incisor area. Mild crowding does not necessarily require the use of superelastic wires, and a classic small diameter work-hardened alloy or a well-established multistranded round stainless steel wire will generally perform as well.

In periodontally compromised patients, and sometimes in the lower incisor area, it would be advisable to maintain the force level delivered to each tooth below 100 g. Data available from properly designed experiments (with 2 mm or more of wire deflection at oral temperature) show that the average delivery force of an austenitic superelastic NiTi .016 × .022-in ranges between 200 g and 300 g. Instead, 35°C and 40°C Thermo-Active Copper Ni-Ti rectangular, Nitinol SE and Nitinol XL, and Neo Sentalloy F240 wires of similar diameters deliver forces around 100 g. In order to obtain lower forces from austenitic NiTi or multi-braided stainless steel, it is necessary to select smaller diameters and abandon the use of rectangular wires during the alignment phase of treatment.

True pseudoelastic behavior generated by torquing forces at nominal oral temperature has not been demonstrated, even in copper NiTi alloys and with a considerable increase of the twist. Only with a lowering of the temperature, with cold rinses for example, can the baseline torque be consistently reduced; in 40°C Thermo-Active Copper Ni-Ti the delivery force can be dropped to 200 g/mm for a less than transient time interval. True thermoelastic alloys may therefore be indicated for early torque control during the alignment phase of treatment and in periodontally compromised patients.

Randomized clinical trials, at least those conducted with austenitic NiTi, on the rate of tooth movement and pain experienced, failed to demonstrate a significantly better performance of superelastic wires compared with conventional alloys, such as multistranded stainless steel wires.

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