European Journal of Orthodontics 25 (2003) 417–421

The load/deflection characteristics of thermally activated orthodontic archwires

Farnaz Parvizi and W. P. Rock
School of Dentistry, University of Birmingham, UK

SUMMARY The objective of the study was to investigate the load/deflection characteristics of three commercially available thermally active nickel–titanium orthodontic archwires using a standard nickel–titanium archwire as a control. The thermally active wires were Regency Thermal, Orthoform, and Eurotherm and the control was Memory.

Round 0.4 mm and rectangular 0.4 × 0.56 mm wires were subjected to 2 and 4 mm of deflection in a water bath at temperatures of 20, 30, and 40°C and forces were measured in three-point bend and phantom head situations.

Analysis of variance revealed that, irrespective of the test set up and wire type, wire size had a significant effect ($P < 0.001$) on the forces produced. An increase in size from 0.4 mm round to 0.4 × 0.56 mm rectangular wire approximately doubled the force values for a given deflection. The effect of wire deflection on the force values varied according to the test system, forces being much higher in the phantom head tests than in the beam tests. In the beam tests, an increase in wire deflection from 2 to 4 mm had no significant effect on the forces exerted, but in the phantom head tests the forces produced by each wire at 4 mm deflection were four to five times greater than those at 2 mm deflection.

Each of the thermally active wires produced less force that the non-thermally active wire. However, there was a large variation between the three types of thermally active wire. In the beam tests each 10°C rise in temperature from 20 to 40°C had a highly significant effect on the force produced by each thermoelastic wire ($P < 0.001$). In the phantom head tests there were significant force increases between 20 and 30°C ($P < 0.001$), but between 30 and 40°C the forces did not change significantly.

Introduction

An ideal orthodontic force should produce rapid tooth movement without damage to the teeth or periodontium. It is difficult to be certain of the precise value of such an ideal force as it will depend on the interaction of factors such as tooth size and type of movement (Houston *et al.*, 1996), although there is some agreement that orthodontic forces should lie within the range 0.15–5 N (Rock and Wilson, 1988). Histological studies have shown that continuous optimal force results in direct resorption of the socket wall in areas of pressure within a few days (Reitan, 1967). In order to compensate for tooth movement, the socket wall on the tension side is remodelled by the deposition of new bone.

Excessive force, which exceeds capillary blood pressure, reduces the cellularity of the periodontal ligament so that tooth movement slows down or stops. In extreme cases there may be resorption of tooth roots with associated pulp death (Leach *et al.*, 2001).

The archwire of a fixed appliance is the major component in the alignment of irregular teeth, both vertically and radially. Wires capable of large elastic deflections are popular as they allow greater working ranges and therefore fewer archwire changes (Anusavice, 1996). However, a possible danger associated with the use of highly resilient archwires is that they may be deflected more than is wise in the early stages of treatment in order to engage the brackets of badly displaced teeth. Excessive wire deflection may generate high forces with concomitant damage to the teeth and supporting structures.

Alloys of titanium have become very popular in the aligning phases of orthodontic treatment. The first such alloy, developed by the Naval Ordnance Laboratory during the American space programme, was marketed as Nitinol orthodontic wire, after tests to determine its suitability for use in orthodontics (Andreasen and Hilleman, 1971). There are now three types of nickel–titanium alloy: the original or conventional type, along with superelastic and thermoelastic variants.

Conventional nickel–titanium alloys have a stabilized martensitic structure with 55 per cent nickel and 45 per cent titanium, formed by cold working. Thereafter, the thermodynamic shape memory effect is suppressed so that it is not possible to introduce structural changes by way of temperature alteration (Evans and Durning, 1996). The lack of formability and weldability are disadvantages of nickel–titanium (Kusy, 1997), but otherwise it produces 'springy' wire with linear stiffness so that there is equal loss of force for each fixed increment of deactivation.

Superelastic nickel–titanium is a development of the original stabilized alloy which undergoes a phase transformation from an austenitic to a martensitic crystal structure when placed under stress (Tonner and Waters, 1994). The hysteresis (stress/strain) curve is such that superelastic wires can sustain very large deflections and return to their original shape with the production of moderate and uniform forces. Some superelastic wires contain copper (5–6 per cent) to increase strength and reduce energy loss. Unfortunately these benefits are associated with an increase in the phase transformation temperature above that of the ambient value in the mouth; 0.5 per cent of chromium is therefore added to the alloy to reduce the stress transformation temperature to 27°C.

Thermoelastic alloys exhibit a thermally induced shape/memory effect whereby they undergo structural changes when heated through a transitional temperature range (TTR) (Kusy, 1997). At room temperature the alloy is soft and easily ligated to badly displaced teeth. At mouth temperature the ratio of austenite increases and along with it the stiffness of the wire, so that it more readily attempts to regain the original archform (Bishara *et al.*, 1995). The extent of this effect depends upon the TTR, which can be set specifically by modifying the composition of the alloy or by appropriate heat treatment during manufacture (Buehler and Cross, 1969). Temperature rise through the TTR produces two effects: the slope of the load/deflection plot increases suddenly by a factor between 2 and 4 and the loading and unloading plateaux are raised. Full recovery on unloading may not always be obtained and has been shown to lie in the range 89–94 per cent (Hurst *et al.*, 1990).

Extravagant claims are made by manufacturers for the various types of nickel–titanium alloy and consequently they are very popular for use in the alignment phase of treatment. However, superelastic wires may not always exhibit a clearly discernible deactivation plateau when tested in bending in association with clinically relevant interbracket and deflection distances (Meling and Ødegaard, 1998).

The aim of the present study was to examine the load/deflection behaviour of a number of thermoelastic wires acting as simple beams and when constrained as part of a fixed appliance in a phantom head jaw.

Materials and methods

Examples of three commercially available thermoelastic nickel–titanium archwires were tested in 0.4 mm round and 0.4 × 0.56 mm rectangular sections, using a conventional nickel–titanium product as a control. The thermally active wires were: Regency Thermal (Direct Ortho, Bristol, UK), Orthoform (3M Unitek, Bradford, UK), and Eurotherm (Ortho Care, Bradford, UK). The control was Memory wire (3M Unitek).

The forces were measured using a mechanical testing machine (Instron model 5544, Instron Ltd, High Wycombe, UK) in association with Instron Merlin software. The wires were tested in two situations: first as simple beams and second as part of a fixed appliance on a phantom head jaw. All tests were carried out in a water jacket supplied from a water bath having proportional temperature control to within 1°C (Figures 1 and 2).

Beam tests were carried out using a jig machined from stainless steel and incorporating 1.25 mm steel rods set 15 mm apart to simulate a typical interbracket span (Rock and Wilson, 1988). A 4 cm length of test wire cut from a straight length was positioned centrally on the jig and deflected at its midpoint using a chisel edge attached to the load cell of the Instron. The rectangular wires were placed with the longer dimension flat to simulate the clinical situation. A crosshead speed of 1 mm/minute was used to a maximum deflection of 4 mm. Five specimens of each wire were tested.

For the simulated fixed appliance tests the teeth of a plastic phantom head jaw were fitted with Andrews' prescription 0.55 × 0.7 mm slot straight wire brackets and buccal tubes. Secure attachment was achieved for

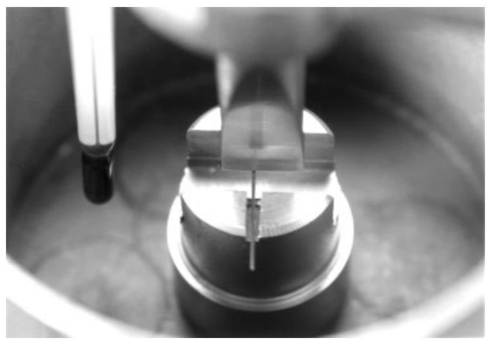

Figure 1 The beam test jig in position in the water jacket (without water).

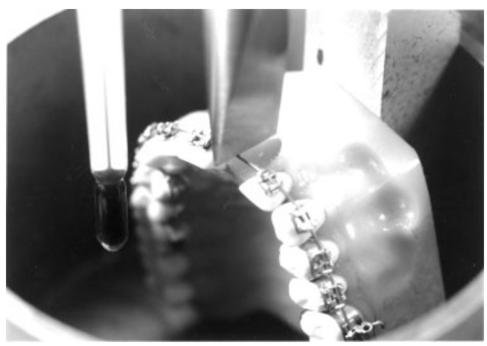

Figure 2 A test in progress on the phantom head jaw (without water).

the brackets by soldering a short length of stainless steel wire to the base of each and cementing this into a hole prepared through the crown. Accurate slot alignment was achieved by using a plain 0.45 × 1.0 mm archwire as a former while the cement set. Tests were carried out using pre-formed archwires in the same water jacket used for the beam tests, by deflecting the wire at the midpoint of a space created by removing the upper left central incisor (Figure 2).

Statistical analyses

The results were analysed using ANOVA according to the General Linear Model program on Minitab version 10 (Minitab Inc., Pennsylvania, USA). Variables for which differences were significant at the $P < 0.05$ level were further investigated using Tukey's pairwise comparisons.

Results

The mean forces exerted by 0.4 mm round wires in the beam test at 2 mm deflection ranged from 0.75–1.32N at 20°C (Table 1). At 40°C the forces ranged from 1.31–1.70N. ANOVA demonstrated significant increases in force levels for all wires as the temperature increased over the test range. Although the 29 per cent force rise with conventional Memory wire was less than the 70 per cent increases for each of the three thermoelastic wires, it was still significant.

Increasing deflection from 2 to 4 mm produced modest increases in force but they were not significant for any wire (Table 1).

An increase in wire cross-section from 0.4 mm round to 0.4 × 0.7 mm rectangular approximately doubled each mean force value from that found in the beam tests with round wires at both 2 and 4 mm of deflection (Table 2).

Statistical analysis of the beam test results using ANOVA to investigate the effects of the main variables, wire type, size, temperature and deflection showed no significant differences between the forces exerted by the three thermoelastic wires. However, each exerted less force than conventional nickel–titanium for a given set of variables according to Tukey's pairwise comparisons.

The forces exerted by 0.4 mm wires in the phantom head were around four times those of comparable beam test values at 2 mm deflection. At 4 mm deflection the phantom head test values were eight times higher (Table 3).

Similar trends were found when 0.4 × 0.7 mm wires were tested. The proportional increases between 2 and 4 mm of wire deflection were again in the region of four and eight times, respectively, for both the beam and the phantom head tests (Table 4).

Discussion

The main aim of the present study was to compare the performance of three thermally activated orthodontic archwires against that of a standard nickel–titanium wire at different temperatures and deflections in a simulated clinical situation. This was undertaken by attaching wires to a fixed orthodontic appliance on a phantom head jaw that was immersed in a temperature-controlled water bath.

In order to assess further the performance of each wire, free from the restraint of being attached to brackets, specimens were also tested as simple beams in three-point bend tests. Kusy and Dilley (1984) recommended three-point bend results as being more reliable than four-point tests as the variances were lower. However, the same authors pointed out that errors may be produced in beam tests due to slippage of the wire inwards across the supports at excessive wire deflections. To ensure that a wire is tested within the range of its metallurgical properties, a deflection of no greater than 5 per cent of span length was recommended. For the present study this would suggest a maximum deflection of 0.75 mm. However, in the mouth wire deflections of between 2 and 4 mm are the norm and these values were therefore used in the present study.

Preliminary beam tests had shown that as deflection increased the measured force rose steadily to a peak at around 4 mm of deflection and then fell off rapidly. This rapid fall was explained by slippage of the wire inwards across the supports as deflection increased, the effect being particularly apparent with 0.4 mm wires

Table 1 Means (standard deviations) of forces measured when 0.4 mm wires were deflected 2 and 4 mm at 20, 30, and 40°C in three-point bend beam tests (N).

	2 mm			4 mm		
	20°C	30°C	40°C	20°C	30°C	40°C
Regency Thermal	0.76 (0.02)	1.03 (0.05)	1.31 (0.05)	0.83 (0.06)	1.13 (0.02)	1.44 (0.10)
Orthoform	0.90 (0.04)	1.14 (0.05)	1.43 (0.11)	1.00 (0.02)	1.20 (0.06)	1.47 (0.09)
Eurotherm	0.75 (0.03)	1.04 (0.03)	1.33 (0.03)	0.79 (0.02)	1.09 (0.04)	1.38 (0.05)
Memory wire	1.32 (0.07)	1.43 (0.04)	1.70 (0.12)	1.58 (0.08)	1.51 (0.12)	1.72 (0.05)

Table 2 Means (standard deviations) of forces measured when 0.4 x 0.7 mm wires were deflected 2 and 4 mm at 20, 30, and 40°C in three-point bend beam tests (N).

	2 mm			4 mm		
	20°C	30°C	40°C	20°C	30°C	40°C
Regency Thermal	1.45 (0.04)	1.87 (0.08)	2.39 (0.11)	1.57 (0.06)	1.98 (0.14)	2.65 (0.03)
Orthoform	1.39 (0.08)	1.94 (0.05)	2.54 (0.05)	1.55 (0.04)	2.12 (0.14)	2.58 (0.09)
Eurotherm	1.53 (0.02)	2.00 (0.04)	2.59 (0.07)	1.63 (0.02)	2.13 (0.08)	2.64 (0.05)
Memory	2.80 (0.14)	3.04 (0.12)	3.26 (0.04)	3.02 (0.30)	3.10 (0.10)	3.35 (0.24)

Table 3 Means (standard deviations) of forces measured when 0.4 mm wires were deflected 2 and 4 mm at 20, 30, and 40°C in a phantom head (N).

	2 mm			4 mm		
	20°C	30°C	40°C	20°C	30°C	40°C
Regency Thermal	3.53 (0.73)	4.16 (0.23)	5.40 (0.43)	7.95 (0.42)	9.63 (1.22)	9.98 (1.10)
Orthoform	4.00 (0.70)	5.41 (1.67)	7.02 (1.15)	11.38 (2.77)	13.70 (4.22)	14.07 (4.25)
Eurotherm	3.00 (0.43)	4.24 (0.39)	4.54 (0.30)	6.60 (0.75)	8.04 (0.57)	7.98 (0.58)
Memory	4.84 (0.33)	5.39 (0.39)	6.22 (0.26)	10.87 (0.83)	11.69 (1.47)	13.51 (1.24)

Table 4 Means (standard deviations) of forces measured when 0.4 x 0.7 mm wires were deflected 2 and 4 mm at 20, 30, and 40°C in a phantom head (N).

	2 mm			4 mm		
	20°C	30°C	40°C	20°C	30°C	40°C
Regency Thermal	7.08 (0.22)	9.31 (0.30)	11.31 (0.39)	15.12 (0.84)	17.15 (0.43)	19.91 (1.52)
Orthoform	6.95 (1.13)	13.50 (1.55)	14.98 (1.21)	18.47 (1.51)	25.17 (1.49)	26.23 (1.58)
Eurotherm	7.31 (1.42)	10.35 (0.70)	11.98 (0.97)	16.22 (1.79)	19.63 (1.32)	20.51 (1.96)
Memory	12.00 (0.81)	14.20 (2.89)	14.70 (1.03)	22.30 (1.30)	24.99 (1.87)	27.94 (0.49)

at deflections above 6 mm. Up to 4 mm deflection the force/deflection curves were linear. Visual observation of the wires revealed no apparent slippage and there was little variation between the forces measured for similar specimens. However, it must be accepted that the wires were deflected far beyond the 0.75 mm limit suggested by the principles of physics and engineering and it would be wrong to interpret the findings of the beam tests other than as between-wire comparisons and as an indication that much higher forces are created when an archwire is deflected against the restraints imposed by incorporation in a fixed appliance.

Beam tests revealed no significant force increases when deflection was increased from 2 to 4 mm. In the phantom head test, increasing deflection significantly influenced the forces produced by all wires. In the beam tests, the three thermoelastic wires behaved similarly and produced lower forces than conventional Memory wire. In the phantom head tests, the thermally active wire, Orthoform, produced significantly higher forces than the other two thermoelastic wires. Indeed the forces exerted by the thermally active wire, Orthoform, and the conventional wire, Memory, supplied by the same manufacturer were similar ($P > 0.05$).

The transition temperatures of some commercial thermoelastic wires lie between 17 and 32°C, but other alloys may not have a suitable transition temperature for them to exhibit superelastic behaviour at body temperature (Yoneyama *et al.*, 1992). In the present study the forces exerted by thermoelastic wires in beam tests increased significantly with each 10°C temperature rise. In phantom head tests, however, although a temperature rise from 20 to 30°C increased forces significantly, further increases between 30 and 40°C were not significant according to ANOVA. Again this was probably due to the restraints imposed on the wire by incorporation in a fixed appliance.

Forces exerted by 0.4 mm wires at 4 mm deflection in the phantom head and all forces recorded for rectangular wires in the phantom head were higher than the 1–2N normally recommended as suitable for orthodontic tooth movement (Graber and Vanardsdall, 1994). At 4 mm deflection in the phantom head forces ranged from 11.3 to 27.9N, 10 times the above recommendation.

Address for correspondence

W. P. Rock
School of Dentistry
St Chad's Queensway
Birmingham B4 6NN, UK

References

Andreasen G F, Hilleman T B 1971 An evaluation of 55 cobalt substituted Nitinol wire for use in orthodontics. Journal of the American Dental Association 82: 1373–1375

Anusavice K J 1996 Phillips' science of dental materials, 10th edn. W B Saunders Company, Philadelphia

Bishara S E, Winterbottom J M, Sulieman A H, Rim K, Jakobsen J R 1995 Comparisons of the thermodynamic properties of three nickel-titanium orthodontic archwires. Angle Orthodontist 65: 117–122

Buehler W J, Cross W B 1969 55-Nitinol, a unique wire alloy with a memory. Wire Journal 2: 41–49

Evans T J W, Durning P 1996 Orthodontic products update: Aligning archwires, the shape of things to come?—A fourth and fifth phase of force delivery. British Journal of Orthodontics 23: 269–275

Graber T M, Vanardsdall R L Jr 1994 Orthodontics. Current principles and techniques, 2nd edn. Mosby, Year Book Inc., St Louis

Houston W J B, Stephens C D, Tulley W J (eds) 1996 A textbook of orthodontics, 2nd edn. Wright, Bristol

Hurst C L, Duncanson M G, Nanda R S, Angolkar P V 1990 An evaluation of the shape-memory phenomenon of nickel-titanium orthodontic wires. American Journal of Orthodontics and Dentofacial Orthopedics 98: 72–76

Kusy R P 1997 A review of contemporary archwires: their properties and characteristics. Angle Orthodontist 67: 197–208

Kusy R P, Dilley G J 1984 Elastic modulus of a triple-stranded stainless steel archwire via three and four point bending. Journal of Dental Research 63: 1232–1240

Leach H A, Ireland A J, Whaites E J 2001 Radiographic diagnosis of root resorption in relation to orthodontics. British Dental Journal 190: 16–22

Meling T R, Ødegaard J 1998 The effect of temperature on the elastic responses to longitudinal torsion of rectangular nickel titanium archwires. Angle Orthodontist 68: 357–368

Reitan K 1967 Clinical and histologic observations on tooth movement during and after orthodontic tooth movement. American Journal of Orthodontics 53: 721–745

Rock W P, Wilson H J 1998 Forces exerted by orthodontic aligning archwires. British Journal of Orthodontics 15: 255–259

Tonner R I M, Waters N E 1994 The characteristics of super-elastic Ni-Ti wires in three-point bending. Part 1: The effect of temperature. European Journal of Orthodontics 16: 409–419

Yoneyama T, Doi H, Hamanaka H 1992 Influence of composition and purity on tensile properties of Ni-Ti alloy castings. Dental Materials Journal 11: 157–164