European Journal of Orthodontics 23 (2001) 233–241

The influence of bracket type on the force delivery of Ni-Ti archwires

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SUMMARY This study investigated the force delivery of an 0.014 inch nickel-titanium (Ni-Ti) archwire used in combination with a range of commercially available bracket systems, and using a model based on an 'ideal' mandibular archform. The model aimed to replicate the clinical interbracket span. The force delivery was measured at four different sites on an archwire for one batch of 10 nickel titanium archwires from one manufacturer, using one bracket/archwire combination. The four sites represented the lateral incisor, canine, second premolar and first molar positions. Force delivery was also measured for a further four different bracket designs at four different sites on the archwire using five fresh wires of the same archwire type. The wires were loaded with an M5 Nene Universal testing machine.

The results demonstrate that the peak and plateau force, both of which are clinically important, are dependent on several factors of the archwire/bracket combination. The results showed that 20 per cent of the batch of 10 wires behaved differently by delivering a higher peak force. There was a statistically significant difference ($P < 0.05$) between the four bracket/archwire combinations for the peak forces delivered, but there was very little difference between the four bracket/archwire unloading force delivery values. The wires delivered a predictable force on the unloading curves, but self-ligating brackets may not develop sufficient strain within the wire to take full advantage of the superelastic effect of Ni-Ti wires.

Introduction

Nickel-titanium (Ni-Ti) superelastic archwires have been used in orthodontic treatment for several years (Waters, 1992; Airoldi and Riva, 1996). Ni-Ti archwires have potential advantages over conventional stainless steel archwires due to their superelastic properties, allowing a suggested constant level of force to be applied to the teeth over a range of displacements. The mechanical properties of Ni-Ti alloys are characterized by the memory effect (Buehler *et al.*, 1963; Andreasen and Morrow, 1978) and the so-called pseudo- or superelastic behaviour (Miura *et al.*, 1986). A stress/strain diagram of a superelastic Ni-Ti alloy demonstrates the well-known non-linear property of a superelastic plateau that is characterized by a wide range of strain with little increase in stress. This superelastic property is due to a phase transformation from body-centred cubic austenitic form to hexagonal close-packed martensitic form when the stress reaches a certain level. The reverse occurs on deactivation. However, unloading the material results in wide hysteresis and it is necessary to distinguish between loading and unloading curves, with the unloading curve at a lower level of stress than the loading curve.

Many investigations of the superelastic properties of Ni-Ti archwires have been reported in the literature (Burstone *et al.*, 1985; Miura *et al.*, 1986; Kusy and Wilson, 1990; Mohlin *et al.*, 1991; Segner and Ibe, 1995). It is of interest to the orthodontist to be able to predict the level of force that a particular archwire will deliver to teeth for the conditions that it will be used under, which might be related, for example, to the particular combination of archwire and bracket type for the deflection required.

The force delivery will depend on several factors in the archwire/bracket combination.

Important properties include the superelastic behaviour of the archwire, the dimensions of the bracket (which will clearly influence the interbracket span) the deflection required, the method of ligation and the frictional forces generated at the archwire/bracket interface (Creekmore, 1976; Kusy, 1991). In this study, an experimental model was constructed, based on an 'ideal' mandibular archform, to investigate the force delivery of a Ni-Ti archwire of a commonly used size in combination with a range of commercially available bracket systems at different locations on the archform.

Materials and methods

Model design

An *in vitro* test model was designed to allow evaluation of the force delivery characteristics of superelastic wires in the mandibular archform. The model consisted of a series of Delrin posts fashioned into an archform to represent a mandibular arch (Figure 1a). Each post was 5 mm in diameter and 31 mm in length, and positioned in holes in the base. The linear distance from the centre of each post is shown in Table 1. A locking mechanism allowed free vertical and rotational movement when open, but held the posts firm when closed. The posts and base were constructed from a plastic with a low coefficient of friction (Delrin, RS Components Ltd, Corby, Northants, UK).

Validation of inter-bracket spans

The clinical relevance of the results will be dependent upon realising the limitations of the model. The model was not representative of the true clinical situation in that the pegs were arranged along an aligned archform with no crowding, tipping, rotations or irregularity of

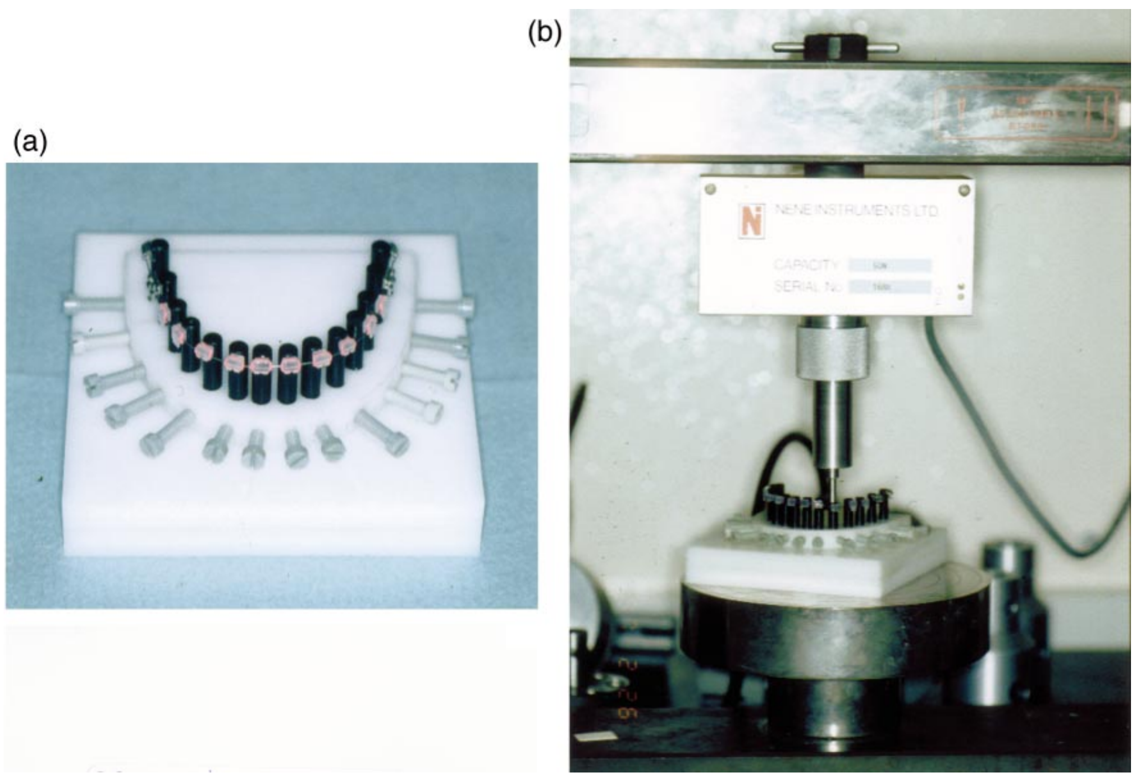

Figure 1 (a) *In vitro* mandibular archform model. (b) Archform model set up for force measurements in the Nene M5 Universal tester.

Table 1 Linear distance (in mm) between the centre of each post in the mandibular archform model.

LL67	LL56	LL45	LL34	LL23	LL12	LL12	LL23	LL34	LL45	LL56	LL67
11.7	11	8.25	8.5	7	6.5	6.5	7	8	8	11	11.5

interbracket spans. Linear measurements were taken of the inter-bracket spans of 10 sets of well aligned orthodontic casts. This demonstrated that the model spans generally overestimated the interbracket spans, especially in the molar region, although the majority fell within 2 standard deviations of the population mean. However, the dental irregularity found *in vivo* varies considerably, and when a span is too small to ligate the archwire fully into the bracket, partial ligation or non-ligation of the tooth into the system is employed. Similarly, space within the arch is often created as teeth are extracted to allow alignment. Therefore the overestimation and the model spans used can be justified and will form a basis on which the clinician can assess force delivery for the 0.014 Ni-Ti archwire.

Method

Two tests were performed:

1. The force delivery was measured at four different sites on an archwire for one archwire/bracket combination for one batch of 10 Ni-Ti archwires from one manufacturer. The four sites represented: canine (site A); second premolar (site B); first molar (site C); and lateral incisor (site D).

The wires were fitted into the model, which was then placed in the tensile tester as shown in Figure 1b. The deflection was chosen to be around 3 mm so that it was within the clinically relevant range observed at initial placement of archwires. The experimental set up used the tensile tester in the time mode that gave an actual maximal deflection of 2.75 mm. Four readings for each site were taken on each of the 10 wires. Average force deflections were then calculated for each wire at each site.

2. The force delivery was measured for four different bracket designs at four different sites on the archwire using five fresh wires of the same archwire type as in test 1. Once again the wires were deflected by 2.75 mm and four repetitions of the readings were taken on each wire at each site. This test included aesthetic brackets, which are commonly used in the upper arch, whereas the model was designed on the lower arch. To overcome this, the first premolar position (site E) was used instead of site D (lateral incisor) to simulate maxillary arch interbracket spans.

The bracket/archwire/ligation systems

The archwires used in this study were all 0.014-inch diameter Ni-Ti superelastic alloys (Forestadent Ltd, Milton Keynes, UK) as received from the manufacturer. The archwires were cut to a constant length so as to allow 2 mm of archwire to protrude from the distal end of the terminal brackets. The terminal extensions of the archwire were cinched back after annealing.

The bracket systems used were 0.022 inch × 0.028 inch pre-torqued and pre-angulated brackets as follows.

For test 1:

1. Andrews lower arch prescription straight wire appliance, non-extraction series (A-Company Orthodontics, San Diego, CA, USA). This is a stainless steel bracket of Siamese twin design. The mesio-distal widths measured 3.0 mm for the lower incisor brackets and 3.5 mm for the canine and premolar brackets.

For test 2:

1. Mini Uni-Twin, upper arch Roth prescription. (3M Unitek, CA, USA). This is a stainless steel Siamese twin bracket. The mesio-distal widths measured 3.0 mm for the maxillary lateral incisor, and first and second premolars. The maxillary central incisor and canine measured 3.5 mm.
2. Transcend ceramic bracket, upper arch Roth prescription (3M Unitek). This bracket has a single twin design. The mesio-distal width measured 3.5 mm.

3. Clarity metal reinforced ceramic bracket (3M Unitek). The upper arch Roth prescription was used. The design is of the Siamese twin variety. The mesio-distal widths were 3.5 mm for the maxillary incisors and 4.0 mm for the maxillary canines and premolars.
4. Damon stainless steel self-ligating bracket (A-Company). The Andrews upper arch prescription was used. The mesio-distal dimensions were 3.0 mm for the maxillary lateral incisors, and 3.5 mm for the maxillary canines and premolars.

The wires were ligated with a fresh standard alastik (3M Unitek) after each archwire change, with the exception of the Damon bracket series, according to a standardized ligation method for comparative purposes.

The brackets were bonded to the posts, taking care to control alignment without the use of positioning jigs. Horizontal alignment was controlled by ensuring the tie wings were parallel to the long axis of the posts. The vertical alignment was fixed by measurement of the centre of the slot to the top of the posts, and rotation errors were controlled by measurement of either end of the slot to the top of the post. The in/out variations in position were controlled by the accuracy of the machining of the model.

Loading

The wires were loaded with an M5 Nene Universal testing machine (Deltalab-Nene Ltd, Nottingham, UK). The Davenport Nene materials test program was used. All measurements were performed dry. The wires were loaded by an aluminium probe that had the same dimensions as the plastic posts (Figure 1b). The load cell used had a maximum load capacity of 50 N. The deflection rate was set at 5 mm/minute in both tests. The maximum deflection was 2.75 mm.

Measurements of force within the wire were recorded by the load cell for the deflections of the posts in the vertical dimension. The error limits for measurement of stress within the wire lay within the limits of +0.19 to –1.05 per cent. The distance error limits of the load cell were velocity-dependent and were calculated at 10 per cent. A separate experiment was run to evaluate the systematic error. This was investigated by measuring the performance of a new wire and repeating the tests with the same wire 2 weeks later.

The data were collected in the form of ASCII files by the Davenport Nene materials test program, which recorded force of the sample against time. The time interval for a particular force was then multiplied by the deflection rate of 5 mm/minute to calculate the force/deflection data. This was carried out for the loading and unloading cycle, which took a total of 66 seconds to complete. The data were processed and analysed using the Microsoft Excel spreadsheet program. The ambient temperature was recorded with a standard thermometer and remained within the range of 21–24°C.

Results

The continuous data were sampled to give unloading plateau values at 1.5 mm deflection and peak loading values. The mean, standard deviation and 95 per cent confidence intervals were calculated. The data were tested for normality using Minitab Inc. for Windows (State College, Philadelphia, PA, USA), version 11.2. This revealed that the data were distributed normally. One-way analysis of variance (ANOVA) was used to identify significant differences in force delivery between the groups using the SAS program (SAS Inst. Inc., NC, USA). This was undertaken by grouping values that were statistically different into Waller groupings. The level of significance was set at $P < 0.05$.

Test 1

The force delivered for four repetitions at four different sites on 10 Ni-Ti wires were tested.

Figure 2 presents the data produced at site A. These show that there was an intrabatch variation amongst the 10 wires. Two of the 10 wires delivered more peak force at site A (4.29–4.83 N) than the rest of the wires. The other eight wires delivered a peak force within the relatively narrow boundaries of 3.10–3.61 N. The unloading plateau phase was characterized by a narrow distribution of force (0.67–1.14 N) for all the

10 wires. Intrabatch variation was also observed at sites B, C and D, and appeared to vary from one site to another but with the same general trend, that is that the peak force varied more than the unloading plateau phase.

Figure 3 presents the average force/deflection ratio for the 10 wires at each site. Overall the peak force delivery varied considerably between the maximum for site D (4.82–9.16 N) to the minimum for site C (2.35–4.29 N), with sites A and B (2.6–4.83 N) lying in between. The unloading plateau values for all four sites lay within the narrow band of 0.16–1.14 N and were therefore much less interbracket span-dependent.

Statistical analysis showed that there were three statistically significant groupings of wires for site B ($P < 0.05$) on peak loading, whereas there was only one on the unloading plateau value. There were between four and five statistically significant groupings for peak loading and unloading plateau values for sites A, C and D. The differences in the unloading plateau values, however, were very small between groups of wires.

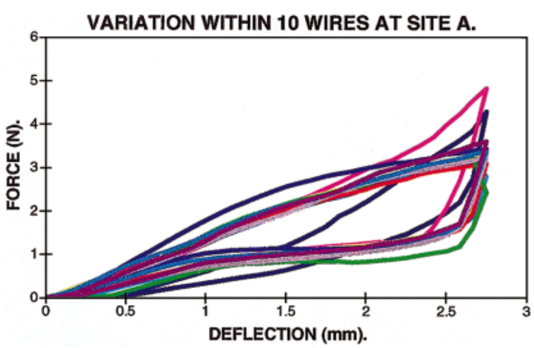

Figure 2 Graph showing the variation in force/deflection relationship for a batch of 10 nickel titanium wires at site A (canine).

Test 2

Figure 4 illustrates the force delivery at site A for the four bracket types. This shows that the peak force at site A was greatest for the Transcend bracket (7.5 N), similar for the Clarity (3.8 N) and Mini Uni-Twin brackets (4.4 N), and least for the Damon brackets (2.5 N). The unloading values were the reverse with the plateau value (1.1 N) for the Damon bracket being larger than the others, which fell into the narrow range of 0.3–0.5 N.

The force delivery at site B for the four bracket types is shown in Figure 5. The peak force delivery at site B were Transcend (3.8 N), Clarity (2.9 N), Damon (2.9 N) and Mini Uni-Twin (2.6 N). The unloading plateau values were closely grouped between 0.8 and 1.1 N. The peak forces were lower than those at site A, but the unloading plateau values were higher.

Figure 6 demonstrates the force delivery at site C for the four bracket types. Site C showed the following distribution for peak values: Transcend, 3.1 N; Clarity, 2.4 N; Mini Uni-Twin, 2.3 N; and Damon, 2.2 N. The unloading plateau values were all closely grouped in the range 0.8–0.9 N. The overall generation of force was very similar to that at site B.

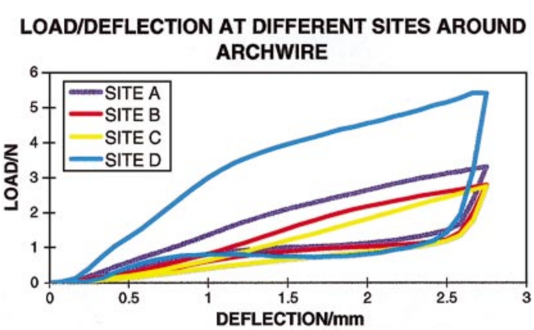

Figure 3 The average force/deflection relationship for the 10 wires at each site: site A (canine), site B (second premolar), site C (first molar) and site D (lateral incisor).

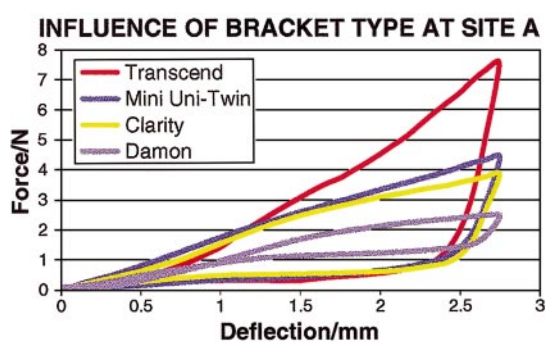

Figure 4 The average force /deflection relationship for the four bracket types at site A (canine).

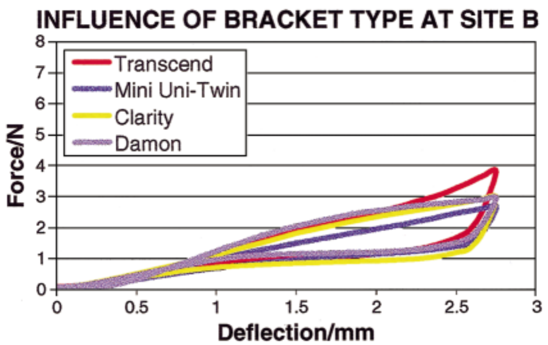

Figure 5 The average force/deflection relationship for the four bracket types at site B (second premolar).

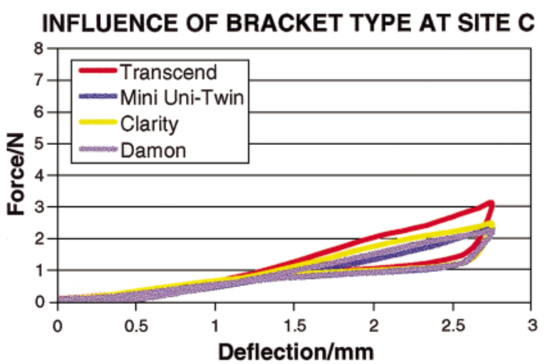

Figure 6 The average force/deflection relationship for the four bracket types at site C (first molar).

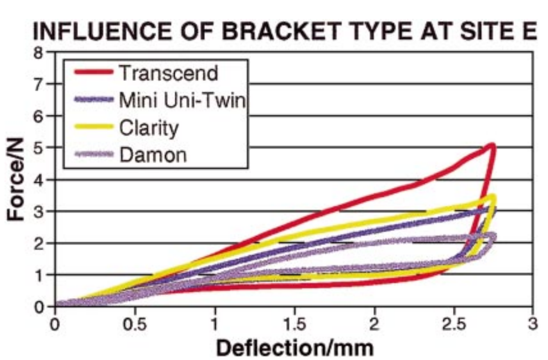

Figure 7 The average force/deflection relationship for the four bracket types at site E (first premolar).

The peak force values at site E for the four bracket types were: Transcend, 5.0 N; Clarity, 3.4 N; Mini Uni-Twin, 3.0 N; and Damon, 2.2 N (Figure 7). Surprisingly, these were slightly greater for Transcend and Clarity at the second premolar, and the order changed slightly but was consistent with interbracket span difference for sites B and E (see Table 1). The unloading plateau values fell within the range 0.6–1.1 N.

The loading curves for the Transcend, Clarity and Mini Uni-Twin brackets did not demonstrate clear plateau regions. However, the unloading plateau for all the bracket types demonstrated a stable force value over deflections of around 0.5–2.0 mm. A more consistent plateau region was observed for the Damon brackets on the loading curve at all sites. However, the plateau region was beyond the clinically useful range.

Statistical analysis (Table 2) demonstrated that at all the sites the Transcend brackets gave a statistically significantly higher peak force than all the other brackets. In each case the peak load for the Mini Uni-Twin and Clarity brackets was not statistically significantly different. At sites A and E, the peak load for the Damon bracket was statistically significantly lower than the others ($P < 0.05$), but was the same as the Clarity and Mini Uni-Twin at sites B and C. In all cases the unloading plateau values fell within a narrow range, although statistically significant differences were observed (Table 2). In particular, the Damon bracket had a statistically higher unloading plateau value at sites A, B and E than the other brackets.

The random error was calculated by the standard error of the differences of two sets of 25 measurements of force delivery at 2.75 mm of wire deflection. The same wire/bracket system was used for both sets of measurements. The variance of the differences was 0.008 N. The standard error attributable to one set of measurements was 0.06 N. The variance of the measured variable for the 25 measurements equalled 0.58 N. The coefficient of reliability was therefore 0.986.

The systematic error was calculated by carrying out a *t*-test on the two sets of 25 measurements. The result was highly significant ($P < 0.01$), but the difference between the sample means was small (0.07 N).

Table 2 The average peak load and unloading plateau (in N) values for each bracket at each site with the Waller groupings, demonstrating statistical significance at the $P < 0.05$ level.

Site	Bracket	Average peak load	Waller* grouping	Average unload plateau	Waller* grouping
A	Transcend	7.5	A	0.3	C
A	Mini Uni-Twin	4.4	B	0.4	C B
A	Clarity	3.8	B	0.5	B
A	Damon	2.5	C	1.1	A
B	Transcend	3.8	A	1.0	B
B	Mini Uni-Twin	2.6	B	1.0	B
B	Clarity	2.9	B	0.8	C
B	Damon	2.9	B	1.1	A
C	Transcend	3.1	A	0.9	A
C	Mini Uni-Twin	2.3	B	0.8	B
C	Clarity	2.4	B	0.9	A
C	Damon	2.2	B	0.8	B
E	Transcend	5.0	A	0.6	C
E	Mini Uni-Twin	3.0	B	0.9	B
E	Clarity	3.4	B	0.9	B
E	Damon	2.2	C	1.1	A

*Waller groupings denoted by a different letter were significantly different.

Discussion

It would be very beneficial to orthodontists to be able to predict the force delivery of an orthodontic archwire in combination with a particular bracket system. Certain variables may influence the force delivery of Ni-Ti archwires, and in this study the range of variation in the force delivery within a commercial batch of 10 similar archwires was investigated and thus their superelastic behaviour was in the clinically relevant range of force and deflection. The variation in force delivery at the different positions on the archwire for different bracket types was explored. This investigated the influence of interbracket span and frictional behaviour at the bracket/archwire interface.

The intrabatch variation will be influenced by the quality of manufacture, in particular alloy composition and impurities, and the interbracket distance by a number of factors, including the design of the bracket, and tooth size and the degree of dental irregularity. The force delivered will depend upon the wire deflection, the size of the span and the superelasticity of the wire. The bracket material will affect the bracket/archwire surface interaction. This in turn will affect the friction between the wire and bracket, and thus the force delivered to the tooth (Kusy, 1991).

Segner and Ibe (1995) noted that only four out of 16 0.014-inch Ni-Ti wires described by their manufacturer as superelastic truly behaved as such in clinically relevant force and deflection ranges. The results from test 1 in this study demonstrated that the majority of the wires behaved superelastically but that up to two out of 10 wires did not show true superelastic behaviour in that a clear plateau region was not observed on the loading curve. On the unloading curve, however, there was a clear plateau region for all wires. On a clinical level, Storey and Smith (1952) found quicker initial canine movement with light loads (250 g) in comparison with heavy loads (500 g). The difference in peak force delivered can be as high as 2.0 N, depending on the site on the archwire that the deflection is required, and is therefore of the same order as the difference between 'light' and 'heavy' forces. This level of intrabatch variation may have a clinical significance. This may be even more critical if heavier rectangular Ni-Ti wires are used.

In this investigation the unloading curves fell within narrower limits, although the degree of superelastic behaviour varied. The unloading plateau occurs at clinically useful displacements between 1 and 2.5 mm. Nine out of 10 wires delivered 0.7–1.0 N of force at 1.5 mm of deflection at all sites. These forces correspond to the values reported by Proffit (1993) for tipping movements, rotations and extrusions (50–70 g). These are the movements that predominate during levelling and aligning with the 0.014-inch Ni-Ti wire. The unloading plateau region is the force value most likely to be applied in the clinical situation as soon as some movement of the teeth has occurred within the periodontal ligament. In this position very little intrabatch variation was observed and also there appeared to be very little influence in the force delivery at the different sites on the dental arch. This suggests that interbracket distance has very little influence on force delivery. It can be concluded therefore that this wire is operating within a clinically relevant range when twinned with the Andrews prescription pre-torqued bracket.

The situation changes, however, when different bracket systems are used. The peak forces tended to be larger with Transcend, Clarity and Mini Uni-Twin, at 7.54, 3.8 and 4.4 N, respectively, than for Damon at 2.5 N, for site A. The unloading curves fell within the narrow limits at site A of 0.3–0.5 N for Transcend, Mini Uni-Twin and Clarity brackets. These values would be slightly lower than the optimal limits quoted by Proffit (1993), although the other sites tested scored more favourably. The use of Damon brackets conversely scored an unloading force above the suggested limits at site A. If one considers the reduced friction associated with these self-ligating brackets, there is the potential to apply too great a force for optimal tooth movement.

In this study, the wires were tested in dry conditions. The effects of salivary lubrication on friction is controversial. Kusy *et al.* (1991) and Kusy and Whitley (1992) have stated that experiments carried out in artificial saliva are invalid as they are not an adequate replacement for human saliva, although Andreasen and Quevedo (1970) felt that saliva played an insignificant role. Read-Ward *et al.* (1997) concluded that the presence of human saliva had an inconsistent effect on static friction and sliding mechanics. Sometimes the saliva acted as a lubricant and at other times it increased friction. On balance it was therefore decided to test the wires dry.

Lundgren *et al.* (1996) has shown that the action of the oral musculature results in a 20 per cent reduction in archwire force over 3 days. This action has not been reproduced in this model. The results may therefore be applicable when the wire is first tied in, but will become less so the longer the wire remains *in situ*.

The errors in the measurements have been described above as being due to random and systematic errors. The largest element of measurement error will be represented by the variation in velocity, and thus distance, recorded by the load cell (10 per cent). This would result in a variation in deflection and thus force delivered by the wire. This size of error will have little effect on the clinically important unloading plateau as 5 per cent deflection either side of 1.5 mm deflection will still be on the plateau. It will, however, affect the peak force values. This is unlikely to be clinically significant as the viscoelastic effect of the periodontal ligament will absorb this force and as soon as the tooth moves within its ligament, it will be on the unloading curve.

It must be recognized that the optimal force range is a difficult value to define in rigid terms and that it is a biological fact that different teeth will respond at different speeds and at different loading. It is therefore the responsibility of the clinician to adapt their technique and procedures to the patient's biological response.

Human error in setting up the model has not been assessed and therefore the contribution to the variation in force delivery is unknown. This error would affect bracket positioning, ligation and wire placement. Attempts were made to minimize these by positioning all the brackets along the peg long-axis flush with the peg end. All the wires were cut to the same length and new alastick ligatures were used for each test.

Conclusions

Twenty per cent out of the batch of 10 wires behaved differently by delivering a higher peak force. The wires delivered a predictable force on unloading.

Self-ligating brackets may not develop sufficient strain within the wire to take advantage of the superelastic effect.

Notwithstanding the different peak forces produced with the four different bracket types, there was very little difference between the unloading force delivery values for the four brackets.

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References

Airoldi G, Riva G 1996 Innovative materials: the NiTi alloys in orthodontics. Bio-Medical Materials and Engineering 6: 299–305

Andreasen G F, Quevedo F R 1970 Evaluation of frictional forces in the 022 × 028 edgewise brackets *in vitro*. Journal of Biomechanics 3: 151–160

Andreasen G F, Morrow R E 1978 Laboratory and clinical analyses of nitinol wire. American Journal of Orthodontics 73: 142–151

Buehler W J, Gilfrich J V, Wiley R C 1963 Effect of low-temperature phase changes on the mechanical properties of alloys near composition of TiNi. Journal of Applied Physics 34: 1475–1484

Burstone C J, Qin B, Morton J Y 1985 Chinese NiTi wire A new orthodontic alloy. American Journal Of Orthodontics 87: 445–452

Creekmore T D 1976 The importance of interbracket width in orthodontic tooth movement. Journal of Clinical Orthodontics 10: 530–534

Kusy R P 1991 Materials and appliances in orthodontics: brackets, archwires and friction. Current Opinion in Dentistry 1: 634–644

Kusy R P, Whitley J Q 1992 Influence of fluid media on the frictional coefficients in orthodontic sliding. Journal of Dental Research 71: 168 (Abstract)

Kusy R P, Whitley J Q, Prewitt M J 1991 Comparison of the frictional coefficients for selected archwire bracket slot combinations in the dry and wet states. Angle Orthodontist 61: 293–302

Kusy R P, Wilson T W 1990 Dynamic mechanical properties of straight titanium alloy arch wires. Dental Materials 6: 228–236

Lundgren D, Owman-Moll P, Kurol J, Martensson B 1996 Accuracy of orthodontic force and tooth movement measurements. British Journal of Orthodontics 23: 241–248

Miura F, Mogi M, Okura Y, Hammanaka H 1986 The superelastic property of Japanese NiTi wire for use in orthodontics. American Journal of Orthodontics and Dentofacial Orthopedics 90: 1–10

Mohlin B, Müller H, Ödman J, Thilander B 1991 Examination of Chinese NiTi wire by a combined clinical and laboratory approach. European Journal of Orthodontics 13: 386–391

Proffit W R 1993 Contemporary orthodontics. Mosby Year Book Inc, St Louis

Read-Ward G E, Jones S P, Davies E H 1997 A comparison of self ligating and conventional orthodontic bracket systems. British Journal of Orthodontics 24: 309–317

Segner D, Ibe D 1995 Properties of superelastic wires and their relevance to orthodontic treatment. European Journal of Orthodontics 17: 395–402

Storey E, Smith R 1952 Force in orthodontics and its relation to tooth movement. Australian Dental Journal 56: 11–18

Waters N E 1992 Superelastic nickel titanium wires. British Journal of Orthodontics 19: 319–322