Original Article

Nickel titanium T-loop wire dimensions for en masse retraction

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ABSTRACT

Objective: To compare the force system produced by nickel-titanium T-loop springs made with wires of different dimensions.

Material and Methods: Thirty compound T-loop springs were divided into three groups according to the dimensions of the nickel-titanium wire used for its design: 0.016" × 0.022", 0.017" × 0.025", and 0.018" × 0.025". The loops were tested on the Orthodontic Force Tester machine at an interbracket distance of 23 mm and activated 9 mm. The force in the y-axis and the moment in the x-axis were registered while the calculated moment to force ratio was recorded at each .5 mm of deactivation. The data were analyzed by three analyses of variance of repeated measures to detect differences and interactions between deactivation and wire size on force, moment, and moment-force ratios (M/F).

Results: All groups had significantly different forces ($P < .001$). The 0.016" × 0.022" wire produced 1.78N of force while the 0.017" × 0.025" and the 0.018" × 0.025" produced 2.81 N and 3.25 N, respectively. The 0.016" × 0.022" wire produced lower moments (11.6 Nmm) than the 0.017" × 0.025" and 0.018" × 0.025" wires, which produced similar moments (13.9 Nmm and 14.4Nmm, respectively). The M/F produced was different for all groups; 0.016" × 0.022" T-loops produced 6.7 mm while the 0.017" × 0.025" and 0.018" × 0.025" T-loops produced 5.0 mm and 4.5 mm, respectively. An interaction was detected for all variables between deactivation and groups.

Conclusion: The larger wires tested produced higher forces with slight increase on the moments, but the M/F produced by the 0.016" × 0.022" wire was the highest found. (*Angle Orthod.* 2016;86:810–817.)

KEY WORDS: T-loop; NiTi; Orthodontics

INTRODUCTION

Among the existing designs of springs used for retraction, the T-loop spring (T-loop) made from beta titanium alloy is considered to be one of the best because it provides medium to high moment-force ratios (M/F) and a low load-deflection rate.[1–4] Despite those favorable characteristics, after 3–4 mm of deactivation[5,6] there is a significant decrease of force produced by the loop, which requires reactivation by the orthodontist or the addition of chain elastics over the loop[4] in order to generate more force.

Nickel-titanium (NiTi) alloys are characterized by the presence of a pseudoelastic plateau in the load-deflection graph during reverse transformation from stress-induced martensitic transformation,[7,8] a property that has been called "superelasticity." The near to constant load deflection rate in that pseudoelastic plateau could improve the problem of the decrease of the force system in the deactivation of T-loops. However, due to the nonlinear nature of strain-based elasticity of superelastic alloys, M/F may also be affected differently in different wire and loop dimensions to an unknown extent compared with alloys with linear elastic moduli.

Even though the use of NiTi T-loops has already been shown in attempts to maintain the force upon

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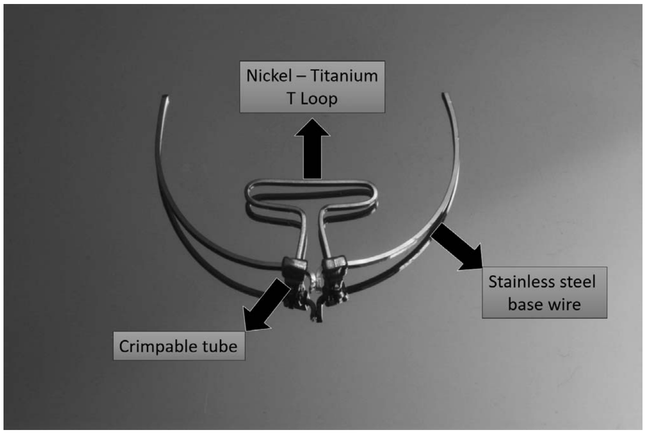

Figure 1. Nickel-titanium composed T-loop. Note that the curved stainless steel base arches would produce the residual moments, while the loop itself would be responsible for producing the forces. Both parts would be involved in producing the moment of activation, while the cross-tubes will protect the region in which they are placed from stress relaxation.

deactivation,[9–12] it appears that they were designed for canine retraction. Because recent studies have shown that there is no difference in the rates of anterior retraction and loss of anchorage between that and en masse retraction,[13–15] it makes sense that the latter method would be clinically more practical and save chair time. The issue is that previously described available NiTi T-loop′ designs produce low forces for that purpose (below 250 gf at the beginning of deactivation), either because they were made from a small dimension wire, such as 0.016" × 0.022",[9] or because they were possibly not activated enough.[10–12] Another issue with the proposed NiTi T-loops is that they are too large, with heights above 8 mm, which could cause impingements clinically.

Therefore, this study aimed to identify the best size of NiTi T-loops to produce a T-loop to deliver forces on the range of 400 gf for en masse retraction and to evaluate the influence of this thickness on the force system produced.

MATERIALS AND METHODS

Thirty superelastic NiTi (Neo Sentalloy, F 100 GAC International, Philadelphia, Pa) T-loops with dimensions of 10 mm length and 6 mm height were produced with a custom device and heat treated at 510°C for 9 minutes.[8,10,12] Then, they were attached to a 0.017" × 0.025" stainless steel base wire through crimpable criss-cross tubes (Morelli Ortodontia, Sorocaba, Brazil) (Figure 1). The springs were distributed into three groups of 10 according to the thickness of the NiTi wire used: 0.016" × 0.022", 0.017" × 0.025", and 0.018" × 0.025".

Felt-tip pen markings were made 7 mm away from the center of the spring toward the end of the base wires in order to align the markings with the entrance of the brackets and ensure a centered positioning of the T-loops. Those measurements allowed the T-loop to be used on a 23-mm interbracket distance (which is the mean distance for segmented arch en masse retraction from molar to canine[16]) with a standardized horizontal activation of 9 mm. Before the base wires were preactivated to a 12-mm radius of curvature,[17] stops were bent on the base wires to ensure that the markings were aligned with the entrance of the brackets attached to the Orthodontic Force Tester, (Chen J, inventor; Apparatus and method for measuring orthodontic force applied by an orthodontic appliance. US patent 6120287 A. September 19, 2000) as illustrated in Figure 2.

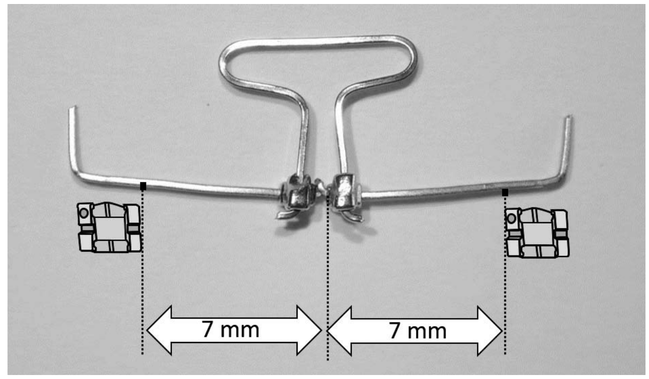

Figure 2. Felt-tip pen markings and stop bends placed on the T-loop. The brackets illustrate the purpose of the stop bend.

The Orthodontic Force Tester was placed inside an isolated acrylic box using gloves; the box was maintained at 37°C by a customized temperature control device. Maxillary self-ligating premolar brackets (0.018" Innovation R, Dentsply GAC International Inc, Philadelphia, Pa) were soldered to the Orthodontic Force Tester′s load cells and separated by 23 mm (Figure 3). The extremities of the base wires were bent so that the marking made earlier matched the entrance of the brackets, while the bend secured the wire in

Figure 3. The orthodontic force tester with a NiTi T-loop secured in place.

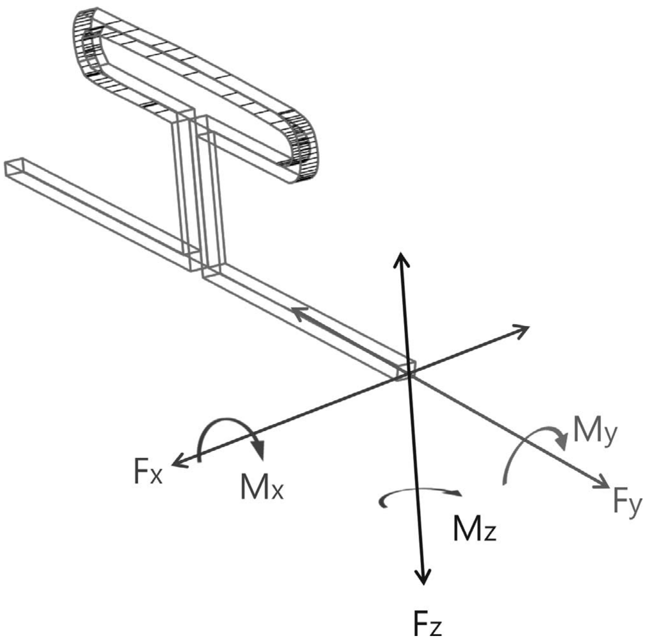

Figure 4. Schematic drawing of forces and moments generated by the T-loop.

place (Figure 2). The produced load system (Fx, Fy, Fz, Mx, My, and Mz) (Figure 4) was recorded at every 0.5-mm spring by the software of the Orthodontic Force Tester and transferred to a Microsoft Excel program (Office 2011, Microsoft, Redmond, Wash). Only Fy and Mx were used in this research as well as the calculated Mx-Fy ratio (M/F).

Since the normality assumption was not violated by the data, two-level analysis of variance of repeated measures was used to detect differences in force, moment, and M/F caused by the wire thickness (groups) and deactivation and to detect possible interactions between these two factors (Table 1). A Tukey post hoc test was used to compare groups and activations for differences. Statistical analysis was performed using SPSS software version 16.0 (Statistical Package for Social Sciences, SPSS Inc, Chicago, Ill).

RESULTS

The sphericity, evaluated by the Mauchly test (which is an analogous test to the homogeneity of the variances done in a one-way analysis of variance) was violated in all tested variables, increasing the chance of a Type I error; thus, the Huynh-Feldt epsilon was used to correct the degrees of freedom and adjust the critical value of force.

The groups produced different profiles of force ($P < .001$). The 0.016” × 0.022” wire produced an average force of 1.78 N, while the 0.017” × 0.025” wire produced 2.81 N and the 0.018” × 0.025” wire produced 3,25 N (Table 1). A significant difference of force was also produced at different activations (Table 2). and an interaction between the groups and activations was detected (Figure 5).

A difference was found in the moments produced by the groups ($P < .001$). The 016” × 0.022” wire produced an average moment of 11.6 Nmm, different from the moments produced by 0.017” × 0.025” and 0.018” × 0.025” wires, 13.9 Nmm and 14.4 Nmm, respectively (Table 3). There was a significant difference

Table 1. Means, Standard Deviations, and Confidence Intervals for Force (N) Between the Groups Over a Range of 7 mm of Deactivation[a]

Group	Mean	SD	95% Confidence Interval Lower Limit	Upper Limit
0.016” × 0.022” (G1)	1.786[A]	.58	1.667	1.904
0.017” × 0.025” (G2)	2.815[B]	.86	2.696	2.934
0.018” × 0.025” (G3)	3.257[C]	.78	3.138	3.376

[a] To transform N into gf, multiply the given values by 102. Different letters attached to the means indicate differences among groups.

Table 2. Means, Standard Deviations, and Confidence Intervals for Force (N) Over a Range of 7 mm of Deactivation[a]

Deactivation	Mean	SD	95% Confidence Interval Lower Limit	Upper Limit
23	4.05[A]	0.88	3.981	4.128
22.5	3.77[B]	0.84	3.704	3.851
22	3.48[C]	0.79	3.413	3.560
21.5	3.22[D]	0.74	3.149	3.296
21	3.00[E]	0.71	2.933	3.080
20.5	2.80[F]	0.67	2.731	2.878
20	2.61[GH]	0.64	2.538	2.685
19.5	2.43[HI]	0.62	2.363	2.510
19	2.29[IJ]	0.60	2.217	2.364
18.5	2.15[JL]	0.58	2.082	2.229
18	2.05[LMN]	0.57	1.982	2.129
17.5	1.96[MNO]	0.56	1.893	2.040
17	1.89[NOP]	0.55	1.822	1.970
16.5	1.82[OPQ]	0.54	1.751	1.898
16	1.75[PQR]	0.52	1.686	1.833

[a] To transform N into gf, multiply values by 102. Different letters attached to the means indicate differences among groups.

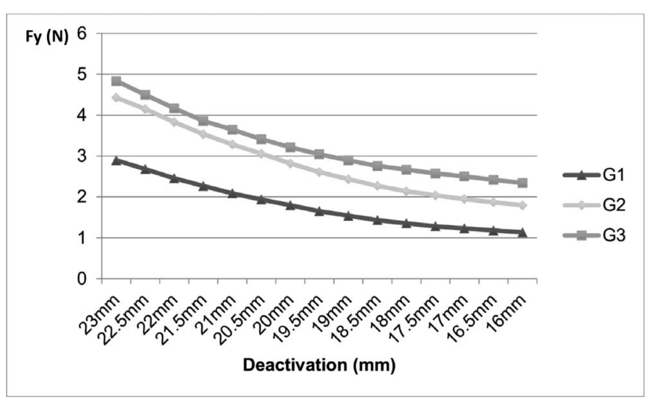

Figure 5. Horizontal force variation in the T-loops over a range of 7 mm of deactivation.

Table 3. Means, Standard Deviation and Confidence Intervals for Moment (Nmm) Between the Groups over a Range of 7 mm of Deactivation[a]

Group	Mean	SD	95% Confidence Interval Lower Limit	95% Confidence Interval Upper Limit
0.016" × 0.022" (G1)	11.64[A]	2.93	10.841	12.445
0.017" × 0.025" (G2)	13.99[B]	3.62	13.196	14.799
0.018" × 0.025" (G3)	14.45[B]	2.97	13.648	15.251

[a] Different letters attached to the means indicate differences among groups.

in the moments produced at different activations (Table 4), and an interaction between the two variables was detected (Figure 6).

There was a difference in the M/F produced by the three wires ($P < .001$), and all groups were different from each other. Loops made with 0.016" × 0.022" wire produced an M/F profile of 6.7 mm, the loops made with 0.017" × 0.025" wire produced 5.0 mm, and the loops made with 018" × 0.025" wire produced 4.5 mm (Table 5). There was a significant difference between activations (Table 6), and an interaction between the two variables was detected (Figure 7).

DISCUSSION

The force produced was different for all groups and force was directly related to wire size. This was expected since it has already been shown that beta-titanium T-loops produced with larger wire sizes produced more force.[18] The NiTi T-loops previously shown in the literature produced lower force levels[10–12] than was found in our study but were probably adequate for canine retraction. This study, however, aimed to evaluate springs for en masse retraction,

Table 4. Means, Standard Deviations, and Confidence Intervals for Moment (Nmm) Over a Range of 7 mm of Deactivation[a]

Deactivation	Mean	SD	95% Confidence Interval Lower Limit	95% Confidence Interval Upper Limit
23	18.45[AB]	2.30	17.939	18.976
22.5	17.66[BC]	2.19	17.141	18.178
22	16.81[CD]	2.09	16.300	17.337
21.5	15.94[DE]	2.02	15.425	16.462
21	15.21[EF]	2.02	14.696	15.733
20.5	14.40[FG]	2.0	13.881	14.918
20	13.73[GH]	1.86	13.215	14.252
19.5	12.97[HI]	1.77	12.457	13.494
19	12.33[IJL]	1.79	11.812	12.849
18.5	11.70[JLM]	1.78	11.181	12.218
18	11.16[LMN]	1.79	10.650	11.688
17.5	10.64[MNO]	1.74	10.129	11.166
17	10.25[NOP]	1.77	9.740	10.777
16.5	9.80[OPQ]	1.71	9.281	10.319
16	9.35[PQR]	1.72	8.827	9.864

[a] Different letters attached to the means indicate differences among groups.

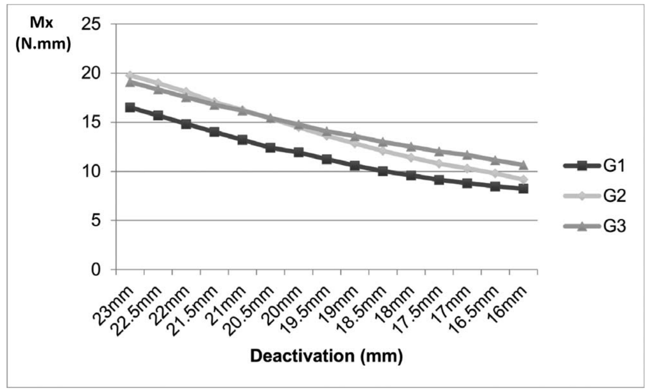

Figure 6. Moment variation in the T-loops over a range of 7 mm of deactivation.

where a force twice as large is generally suggested.[13,14,19] Thus, the 0.016" × 0.022" T-loop produced forces below what is supposedly deemed necessary for en masse retraction (Table 7) but still considered greater than the force deemed necessary for canine retraction.[9] Another T-loop made with the same brand of NiTi we used (Sentalloy, Dentsply GAC International Inc) out of 0.016" × 0.025" wire[9] showed an even smaller force than we did, probably due to its larger height (10 mm) than ours (6 mm) and because of more extensive heat treatment, which decreases the pseudoelastic plateau of NiTi materials.[8] The 0.018" × 0.025" T-loops, on the other hand, delivered forces ranging from 470 gf to 224 gf (Table 7), mostly adequate for en masse retraction and higher than what has been documented in the other T-loops made of the same size wire. These differences are probably due to the height of the T-loop**s** of these reports (8.45 mm[10,12]), which were larger than ours. A peculiar characteristic of shape memory alloys upon reverse transformation from stress-induced martensite is a drop of load before a pseudoelastic plateau is described on a load-deflection graph,[7,20,21] and this effect can be perceived on the forces measured of the T-loops (Table 7; Figure 5). As it has already been suggested for NiTi close coil springs,[22] it may be a good idea to overactivate NiTi T-loops if a more constant force is desired. Compared with a beta-titanium T-loop,[5] where

Table 5. Means, Standard Deviations, and Confidence Intervals for MF Ratio (mm) Between the Groups Over a Range of 7 mm of Deactivation[a]

Group	Mean	SD	95% Confidence Interval Lower Limit	95% Confidence Interval Upper Limit
0.016" × 0.022" (G1)	6.66[A]	0.83	6.399	6.936
0.017" × 0.025" (G2)	5.04[B]	0.35	4.777	5.314
0.018" × 0.025" (G3)	4.50[C]	0.46	4.232	4.769

[a] Different letters attached to the means indicate differences among groups.

Table 6. Means, Standard Deviations, and Confidence Intervals for Moment-Force Ratios (mm) Over a Range of 7 mm of Deactivation[a]

			95% Confidence Interval	
Deactivation	Mean	SD	Lower Limit	Upper Limit
23	4.69[A]	0.81	4.525	4.870
22.5	4.84[AB]	0.84	4.671	5.017
22	5.00[ABC]	0.86	4.827	5.173
21.5	5.12[BCD]	0.87	4.954	5.300
21	5.25[BCDE]	0.89	5.077	5.423
20.5	5.33[CDEF]	0.91	5.157	5.502
20	5.47[DEF]	0.97	5.297	5.642
19.5	5.55[EF]	1.02	5.386	5.731
19	5.62[EF]	1.04	5.450	5.796
18.5	5.68[F]	1.10	5.513	5.859
18	5.71[F]	1.15	5.537	5.882
17.5	5.71[F]	1.20	5.538	5.883
17	5.71[F]	1.24	5.543	5.889
16.5	5.69[F]	1.30	5.522	5.868
16	5.65[EF]	1.40	5.482	5.827

[a] Different letters attached to the means indicate differences among groups.

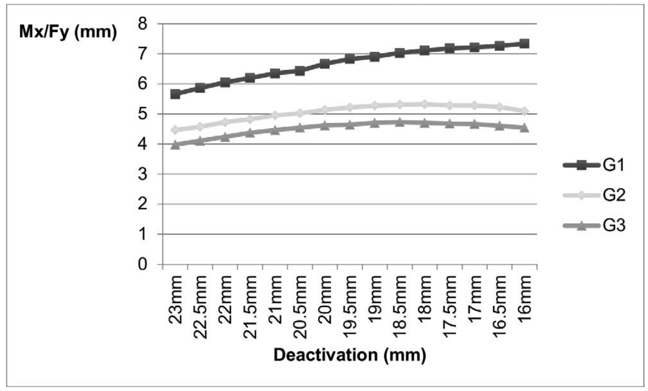

Figure 7. M/F variation in the T-loops over a range of 7 mm of deactivation.

the force produced at 5 mm of activation is approximately 400 gf and drops to around 50 gf at 0.5 mm of deactivation, thus decreasing around 77 gf/mm of deactivation, the 0.018” × 0.025” NiTi T-loop tested has a much more constant deactivation. The fact that it produces around 470 gf at 7 mm of activation and that after the same 4.5 mm of deactivation it produces 270 gf, thus decreasing around 44 gf/mm of deactivation, makes it much more interesting for en masse space closure because it could produce useful forces over a longer span of deactivation. In this study, the T-loops were activated 9 mm, but the data were only collected from the first 7 mm of deactivation because in the last 1 or 2 mm of deactivation the cross-tubes would eventually touch each other and interfere with the deactivation of the T-loops.

There were differences between the groups in the moments produced. The larger wires produced higher moments than the smaller 0.016” × 0.025” wire. This was an expected finding since the moments produced by loops in general are proportional to the size of the wire. Normally, the base wires of T-loops made from beta-titanium wires are preactivated with a curvature around 23 mm of radius to obtain an standard moment.[17] The curvature can be increased or decreased according to the desire of the clinician, decreasing or increasing the moments produced by a particular T-loop.

The challenge in obtaining high moments on a NiTi T-loop comes from the fact that a T-loop has a large stress concentration on its base wire, particularly in the angle between the vertical extensions of the T-loop and the base wires.[23] That large stress concentration induces a stress-induced martensitic transformation that will make that region more flexible rather than more rigid; moreover, further increasing the angulation in that region will cause a different effect than the

Table 7. Means of Generated Forces (N) for Each Group Over a Range of 7 mm of Deactivation[a]

		G1: 0.016″ × 0.022″	G2: 0.017″ × 0.025″	G3: 0.018″ × 0.025″
Variable	Deactivation (mm)	Mean (SD)	Mean (SD)	Mean (SD)
Force	23	2.91 (0.29)	4.42 (0.28)	4.79 (0.19)
	22.5	2.66 (0.27)	4.14 (0.26)	4.44 (0.18)
	22	2.44 (0.24)	3.82 (0.27)	4.11 (0.19)
	21.5	2.24 (0.22)	3.53 (0.28)	3.80 (0.21)
	21	2.06 (0.20)	3.27 (0.26)	3.59 (0.26)
	20.5	1.91 (0.20)	3.05 (0.25)	3.36 (0.24)
	20	1.77 (0.17)	2.81 (0.24)	3.16 (0.23)
	19.5	1.62 (0.16)	2.60 (0.23)	2.99 (0.19)
	19	1.51 (0.15)	2.43 (0.22)	2.84 (0.20)
	18.5	1.40 (0.14)	2.26 (0.21)	2.70 (0.21)
	18	1.32 (0.13)	2.13 (0.20)	2.61 (0.20)
	17.5	1.25 (0.12)	2.03 (0.19)	2.52 (0.21)
	17	1.19 (0.11)	1.94 (0.17)	2.45 (0.22)
	16.5	1.14 (0.11)	1.86 (0.17)	2.36 (0.21)
	16	1.10 (0.10)	1.79 (0.15)	2.29 (0.22)

[a] To transform into gf, multiply by 102.

Table 8. Means of Generated Moments (Nmm) for Each Group Over a Range of 7 mm of Deactivation

		G1: 0.016" × 0.022"	G2: 0.017" × 0.025"	G3: 0.018" × 0.025"
Variable	Deactivation (mm)	Mean (SD)	Mean (SD)	Mean (SD)
Moment	23	16.51 (1.59)	19.85 (2.30)	20.27 (1.71)
	22.5	15.68 (1.55)	19.14 (2.19)	19.37 (1.68)
	22	14.83 (1.52)	18.26 (2.22)	18.53 (1.76)
	21.5	14.01 (1.48)	17.21 (2.25)	17.59 (1.60)
	21	13.21 (1.42)	16.44 (2.28)	16.88 (1.71)
	20.5	12.43 (1.38)	15.56 (2.33)	16.11 (1.72)
	20	11.93 (1.36)	14.65 (2.21)	15.43 (1.74)
	19.5	11.23 (1.32)	13.80 (2.17)	14.71 (1.73)
	19	10.59 (1.29)	13.05 (2.07)	14.13 (1.78)
	18.5	10.02 (1.27)	12.32 (1.90)	13.47 (1.80)
	18	9.58 (1.23)	11.66 (1.94)	13.01 (1.98)
	17.5	9.17 (1.20)	11.17 (1.86)	12.49 (2.05)
	17	8.81 (1.18)	10.69 (1.79)	12.02 (2.20)
	16.5	8.48 (1.15)	10.15 (1.83)	11.47 (2.35)
	16	8.23 (1.12)	9.27 (2.06)	10.90 (2.40)

desired one because higher stresses will make the load/deflection rate of NiTi even lower.[7,21] A solution for that problem has been shown by Bourauel and colleagues[9] with parallel tubes and stainless steel base wires, but that would still allow stress relaxation to occur in that specific region, because there was a 90° bend in the stainless steel base wire after the tubes. We have found a different solution, which was the addition of a crimpable cross tube in that particular area, making it more rigid and adding stainless steel base wires, preactivated by curvature, rather than NiTi base wires, which would probably not produce enough moments. Curvature was added rather than concentrated bends to the base wire because it has been shown that they may be less sensitive to stress relaxation overtime.[6,23]

The M/F produced by the T-loops was different among the groups, which was our most interesting finding since with linear alloys M/F ratios do not appear to change with increased cross section.[18,24] Our results showed similar moments between the two largest wire sizes, but the 0.018" × 0.025" T-loop showed higher forces, thus having a lower M/F ratio (Figures 5 through 7). On the other hand, the 0.016" × 0.025" T-loops, compared with the two larger wire loops, showed much more difference in the moments than the force, thus producing a much higher M/F (Tables 7 through 9). We hypothesize that the most likely explanation for this effect, which should be confirmed in a finite element study, is that the size of the wire affects the stress distribution and the proportion of martensitic transformation differentially as wire size increases, leading to large differences in M/F even though activation is the same. This effect, not previously described, has not been reported as statistically different in alloys that have a linear behavior, such as stainless steel and beta-titanium.[18,24]

The M/F found were insufficient for translatory movement on the larger wires but apparently suitable for controlled tipping[17,25] on the 0.016" × 0.025" wire

Table 9. Means of Moment-Force Ratios (mm) for Each Group Over a Range of 7 mm of Deactivation

		G1: 0.016" × 0.022"	G2: 0.017" × 0.025"	G3: 0.018" × 0.025"
Variable	Deactivation (mm)	Mean (SD)	Mean (SD)	Mean (SD)
MF	23	5.66 (0.44)	4.48 (0.30)	4.24 (0.40)
	22.5	5.86 (0.44)	4.62 (0.38)	4.36 (0.41)
	22	6.05 (0.46)	4.77 (0.44)	4.51 (0.45)
	21.5	6.19 (0.50)	4.87 (0.48)	4.63 (0.45)
	21	6.35 (0.56)	5.01 (0.55)	4.70 (0.47)
	20.5	6.43 (0.63)	5.09 (0.58)	4.80 (0.47)
	20	6.66 (0.64)	5.19 (0.58)	4.88 (0.51)
	19.5	6.82 (0.64)	5.29 (0.60)	4.92 (0.53)
	19	6.89 (0.69)	5.35 (0.58)	4.98 (0.60)
	18.5	7.02 (0.75)	5.41 (0.58)	4.99 (0.66)
	18	7.10 (0.77)	5.45 (0.59)	4.98 (0.71)
	17.5	7.17 (0.76)	5.48 (0.66)	4.96 (0.76)
	17	7.21 (0.85)	5.49 (0.67)	4.90 (0.81)
	16.5	7.26 (0.85)	5.42 (0.71)	4.85 (0.88)
	16	7.33 (0.91)	5.14 (0.94)	4.76 (0.96)

(in which the force was insufficient for anterior retraction) (Table 9). Those values are different from the T-loops already proposed in the literature, which were higher,[10-12] possibly due to their large loop configuration, which may not be as comfortable for patients. Apparently, an increase in the rigidity of the base arch, by increasing its width or further decreasing the radius of preactivation, could solve this issue with this particular T-loop. It is important to point out that this is an in vitro purely mechanical evaluation of the more appropriate wire size to be used in a NiTi T-loop.

The results of this study should be interpreted with care, because they could be misleading. The fact that no translation would be produced with the heavier wires does not mean that this T-loop cannot or should not be used be used clinically. To date, no space closure loop has been able to produce a M/F near 10/1 mm along with a high enough force to retract teeth efficiently. What normally happens when loop mechanics are used is that teeth are first tipped into the space to be closed. From that point on, either the loop is completely deactivated, not producing any horizontal force and therefore bringing the M/F to a high enough value to correct the roots of the teeth, or the moment created on the bracket/base wire interface due to the tipping of the teeth raises the M/F ratio to a higher value, thereby improving control of the roots. Even though the variables tested in this study were not enough to allow a configuration of a T-loop that would produce a high M/F along with a reasonable force for en masse retraction, the data presented allow other variables to be tested in the future, such as preactivation intensity on the base wire or the wire dimension of the base wires.

CONCLUSIONS

This study showed the following conclusions about the springs formed by NiTi T-loops and stainless steel horizontal rods:

- The NiTi 0.017" $\times$ 0.025" and 0.018" $\times$ 0.025" wires seem to be more suitable for en masse retraction.
- The M/F produced for 0.017" $\times$ 0.025" and 0.018" $\times$ 0.025" wires was insufficient for bodily movement or controlled tipping but suitable for tipping when made of 00.016" $\times$ 00.022" wire; however, this force was insufficient for anterior retraction.
- The M/F is dependent on wire size for superelastic NiTi alloys, all other factors being equal.

REFERENCES

1. Faulkner MG, Fuchshuber P, Haberstock D, Mioduchowski A. A parametric study of the force/moment systems produced by T-loop retraction springs. *J Biomech*. 1989;22:637–647.
2. Kuhlberg AJ, Burstone CJ. T-loop position and anchorage control. *Am J Orthod Dentofacial Orthop*. 1997;112:12–18.
3. Menghi C, Planert J, Melsen B. 3-D experimental identification of force systems from orthodontic loops activated for first order corrections. *Angle Orthod*. 1999;69:49–57.
4. Viecilli RF. Self-corrective T-loop design for differential space closure. *Am J Orthod Dentofacial Orthop*. 2006;129: 48–53.
5. Caldas SG, Martins RP, Galvao MR, Vieira CI, Martins LP. Force system evaluation of symmetrical beta-titanium T-loop springs preactivated by curvature and concentrated bends. *Am J Orthod Dentofacial Orthop*. 2011;140:e53–e58.
6. Caldas SG, Martins RP, Viecilli RF, Galvao MR, Martins LP. Effects of stress relaxation in beta-titanium orthodontic loops. *Am J Orthod Dentofacial Orthop*. 2011;140:e85–e92.
7. Burstone CJ, Qin B, Morton JY. Chinese NiTi wire—a new orthodontic alloy. *Am J Orthod*. 1985;87:445–452.
8. Miura F, Mogi M, Ohura Y, Hamanaka H. The super-elastic property of the Japanese NiTi alloy wire for use in orthodontics. *Am J Orthod Dentofacial Orthop*. 1986;90:1–10.
9. Bourauel C, Drescher D, Ebling J, Broome D, Kanarachos A. Superelastic nickel titanium alloy retraction springs—an experimental investigation of force systems. *Eur J Othod*. 1997;19:491–500.
10. Kum M, Quick A, Hood JA, Herbison P. Moment to force ratio characteristics of three Japanese NiTi and TMA dosing loops. *Aust Orthod J*. 2004;20:107–114.
11. Keng FY, Quick AN, Swain MV, Herbison P. A comparison of space closure rates between preactivated nickel-titanium and titanium-molybdenum alloy T-loops: a randomized controlled clinical trial. *Eur J Orthod*. 2012;34:33–38.
12. Rose D, Quick A, Swain M, Herbison P. Moment-to-force characteristics of preactivated nickel-titanium and titanium-molybdenum alloy symmetrical T-loops. *Am J Orthod Dentofacial Orthop*. 2009;135:757–763.
13. Felemban NH, Al-Sulaimani FF, Murshid ZA, Hassan AH. En masse retraction versus two-step retraction of anterior teeth in extraction treatment of bimaxillary protrusion. *J Orthod Sci*. 2013;2:28–37.
14. Heo W, Nahm DS, Baek SH. En masse retraction and two-step retraction of maxillary anterior teeth in adult Class I women. A comparison of anchorage loss. *Angle Orthod*. 2007;77:973–978.
15. Xu TM, Zhang X, Oh HS, Boyd RL, Korn EL, Baumrind S. Randomized clinical trial comparing control of maxillary anchorage with 2 retraction techniques. *Am J Orthod Dentofacial Orthop*. 2010;138:544 e541–e549; discussion 544–545.
16. Martins RP, Buschang PH, Gandini LG Jr. Group A T-loop for differential moment mechanics: an implant study. *Am J Orthod Dentofacial Orthop*. 2009;135:182–189.
17. Burstone CJ, van Steenbergen E, Hanley KJ. *Modern Edgewise Mechanics & The Segmented Arch Technique*. Glendora, Calif: Ormco Co; 1995.
18. Manhartsberger C, Morton JY, Burstone CJ. Space closure in adult patients using the segmented arch technique. *Angle Orthod*. 1989;59:205–210.
19. Sethi P, Saxena R, Keluskar KM. En-masse retraction. *Am J Orthod Dentofacial Orthop*. 2007;132:723; author reply 723–724.
20. Garrec P, Jordan L. Stiffness in bending of a superelastic Ni-Ti orthodontic wire as a function of cross-sectional dimension. *Angle Orthod*. 2004;74:691–696.
21. Vieira CI, Martins RP, Caldas SG, Ribeiro A, Martins LP. Superelasticity and force plateau of nickel-titanium springs. *Dental Press J Orthod*. In press.

22. Tripolt H, Burstone CJ, Bantleon P, Manschiebel W. Force characteristics of nickel-titanium tension coil springs. *Am J Orthod Dentofacial Orthop*. 1999;115:498–507.
23. Junior RS, Caldas SG, Martins LP, Martins RP. Effects of stress relaxation in beta-titanium orthodontic loops: part II. *Angle Orthod*. In press.
24. Faulkner MG, Lipsett AW, el-Rayes K, Haberstock DL. On the use of vertical loops in retraction systems. *Am J Orthod Dentofacial Orthop*. 1991;99:328–336.
25. Tanne K, Koenig HA, Burstone CJ. Moment to force ratios and the center of rotation. *Am J Orthod Dentofacial Orthop*. 1988;94:426–431.