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Effects of stress relaxation in beta-titanium orthodontic loops

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Introduction: This study evaluates the changes in the force system of the beta-titanium T-loop spring (TLS) caused by stress relaxation. **Methods:** Ninety TLSs with dimensions of 6 × 10 mm, made of 0.017 × 0.025-in beta-titanium alloy and preactivated by concentrated bends, were randomly distributed into 9 groups according to the time point of evaluation. Group 1 was tested immediately after spring preactivation and stress relief by trial activation. The other 8 groups were tested after 24, 48, and 72 hours, and 1, 2, 4, 8, and 12 weeks. By using a moment transducer coupled to a digital extensometer indicator adapted to a universal testing machine, the amounts of horizontal forces and moments and the moment-to-force ratios were recorded at every 0.5 mm of deactivation from 5 mm of the initial activation in an interbracket distance of 23 mm. **Results:** The horizontal forces and moments were higher ($P < 0.001$) for group 1 compared with the other 8 groups, which were not different among themselves. All groups produced similar moment-to-force ratios ($P = 0.600$), with no influence of time. **Conclusions:** The TLSs preactivated by concentrated bends had progressive load decreases over time, and this effect is critical in the first 24 hours. (Am J Orthod Dentofacial Orthop 2011;140:e85-e92)

Plastic deformation occurs if the stress exceeds the yield point.[1] What is often overlooked is that plastic deformation can also depend on time.[1,2] Stress-related plastic deformation is referred to as *slip*, and time-related deformation is called *creep*.[1-3] From a microscopic perspective, creep in highly stressed metals is the result of progressive movement of dislocations in the crystalline structure of the material. This microscopic phenomenon can be observed experimentally as an increase in strain associated with constant stress (creep) or a decrease in stress associated with constant strain (stress relaxation) (Fig 1). Creep depends on stress intensity and temperature, because high stresses and temperatures favor the movement of dislocations. In most engineering applications, creep in metals becomes a concern only at temperatures at least 30% of the melting point of the material, because structural components are typically not subjected to high stresses during shaping for a specific application.[1,2]

In orthodontics, straight wires are generally used to construct an appliance. Occasionally, bends or loops are placed in orthodontic archwires to facilitate particular tooth movements. These bends concentrate stress and can cause spacing and unstable dislocations in the crystalline structure at the high-stress points.[1] Orthodontists have tried to overcome this problem with heat treatments in orthodontic appliances made of stainless steel to promote rearrangement of the crystalline structure by relieving residual stresses.[4-6] Another strategy often used is to take advantage of the Bauschinger effect.[7] This consists of overbending the wire and performing several trial activations until the wire assumes the desired shape for the application of the force system. This is important for alloys that are not sensitive to heat treatment, such as beta-titanium (β-Ti).

The β-Ti alloy has a moderate spring-back, somewhere between stainless steel and nickel-titanium alloy,[8,9] and, when used in loops, it might require

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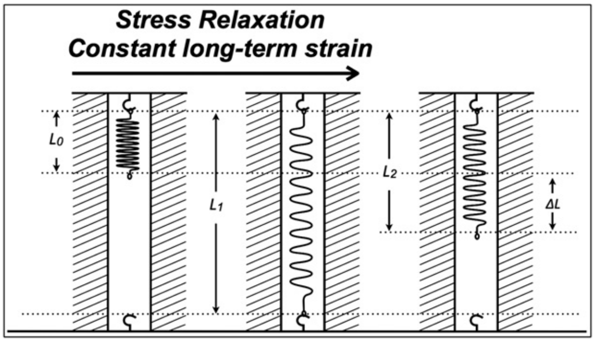

Fig 1. Stress relaxation test in a helical spring. L_o, Original length; L_1, extended length due to activation (constant strain); L_2, total length after stress relaxation; ΔL, deformation due to stress relaxation.

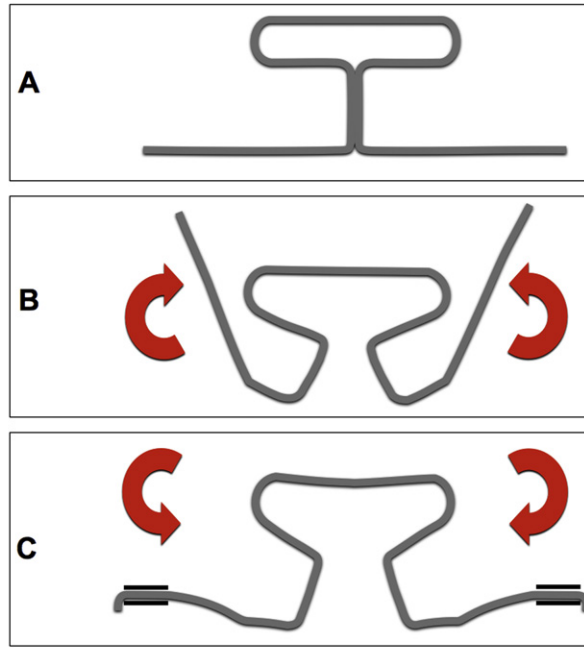

Fig 2. A, TLS in passive form; **B,** TLS preactivated by concentrated bends; **C,** TLS engaged in brackets (loaded in the opposite direction to preactivation).

some stress relief before it can be effectively used.[7,10] The β-Ti T-loop spring (TLS), has been used since the 1980s for space closure, mainly in patients having premolar extractions.[8,11] Templates[7] and other methods[10] of preactivation have been developed to deliver specific forces and enough moment-to-force (MF) ratios, allowing different types of tooth movements. β-Ti TLS stress relief can be done by simulating its activation before placement in the patient's mouth, thereby producing forces in the wire that are suitable for tooth movement.[7,10]

After this, the β-Ti TLS (or any other loop) is loaded in the opposite direction to its preactivation (Fig 2). This subjects the alloy to a constant deformation, which could cause progressive force reduction.[1,2] This time-dependent effect has been thoroughly studied in the alloys used in orthodontics[9,12-16]; however, it has been superficially evaluated for β-Ti, and there are no studies in the literature on this effect with more elaborate geometries, such as in loops.[9,14,16]

Change in the original shape of a loop might prove that contemporary preactivations are not ideal, requiring some adjustments. Thus, the aim of this study was to evaluate the load decay on the force system of TLSs preactivated by concentrated bends over time.

MATERIAL AND METHODS

Ninety TLSs were hand-bent by an author (S.G.F.R.C.) using Marcotte pliers (Hu-Friedy, Chicago, Ill) of 0.017 × 0.025-in β-Ti wires (TMA; Ormco, Glendora, Calif), and a custom template (Fig 3, *A*). The TLSs had dimensions of 6 mm in height by 10 mm in length and were preactivated with concentrated bends (Fig 3, *B*).[10]

The TLSs were randomly divided into 9 groups according to the time of evaluation. Group 1 was tested immediately after spring preactivation and stress relief by trial activation. The other 8 groups had the same procedure but were tested after they were maintained at a 5-mm activation for different times in an interbracket distance of 23 mm. A custom device was specifically made for this purpose (Fig 4). Groups 2 through 9 were kept activated for 24, 48, and 72 hours, and 1, 2, 4, 8, and 12 weeks, respectively.

A universal testing machine (EMIC, São José dos Pinhais, Brazil), equipped with a load cell of 0.1 kN, was coupled to a moment transducer and a digital extensometer indicator (Transdutec, São Paulo, Brazil) for the tests. The speed used for the test was 5 mm per minute, and the digital extensometer's excitation and sensitivity were 5 V and 0.5 mV/V, respectively.

For the tests, the TLSs were positioned symmetrically with an interbracket distance of 23 mm. To ensure the correct activation and the centralization of the TLSs, 9 mm were measured from the center of the loop toward each extremity of the horizontal extensions, where the stops were made (Fig 5).

After an activation of 5 mm, the amounts of horizontal force and moment developed were recorded for every 0.5 mm of deactivation, and the MF ratios were calculated. Furthermore, the amount of horizontal overlap of the vertical extensions of the TLSs in "neutral position" (deformation assumed by a spring when its extremities are placed parallel to the positions that they

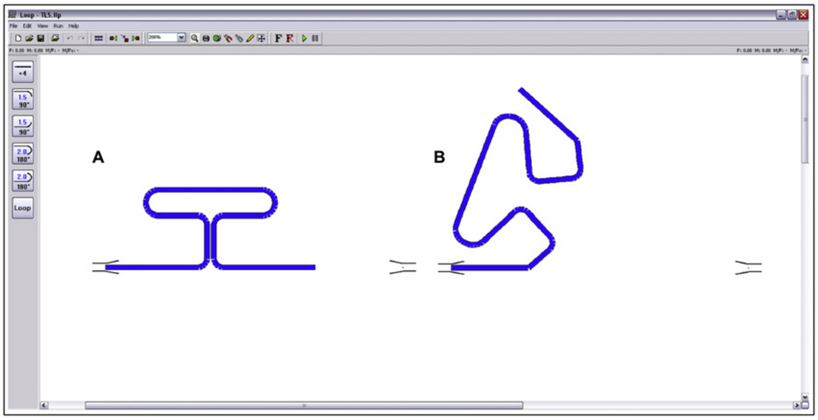

Fig 3. Templates developed in the Loop software (dHAL Orthodontic Software; Kifissia, Greece): **A,** for bending the TLSs; **B,** for the preactivation of the TLSs by concentrated bends.

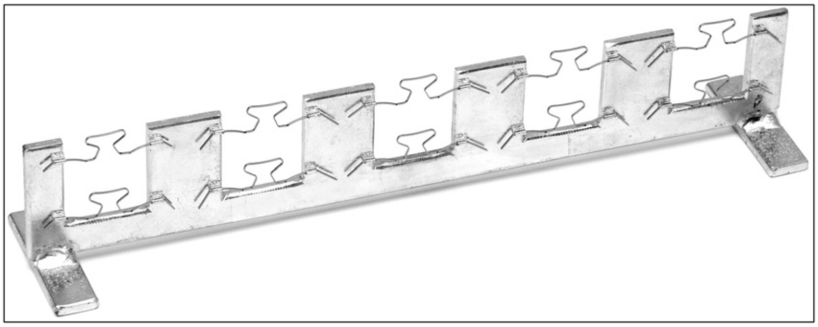

Fig 4. Custom-made device to keep the TLSs activated at 5 mm.

will be engaged on the patient's brackets, not producing any force, only moments) was calculated by linear interpolation. The load-deflection (LD) ratio (slope of the deactivation graph) was also obtained based on the chart (Fig 6).

Statistical analysis

SPSS statistical analysis software (version 16.0; SPSS, Chicago, Ill) was used in this study. The Kolmogorov-Smirnov test indicated normal distributions, and the Levene test showed that all variables had similar variances, except the MF ratios.

The multivariate profile analysis with the procedure for analysis of repeated measures was used to detect differences in forces, moments, and MF ratios among the groups. This analysis compared the total profile, or deactivation pattern, of a whole group in relation to time and deactivation. To identify the differences among the groups, the post-hoc Tukey test was used with the averages generated by each time (total profile average).

Analysis of variance was used, at a level of 5%, to detect differences among the groups in the LD ratio and the amount of overlapping of the vertical extensions of the

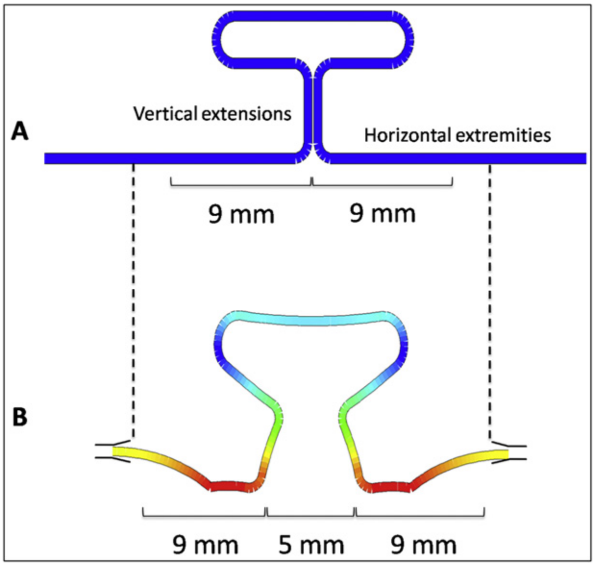

Fig 5. **A,** Loop regions and dimensions on the horizontal extremities used to ensure the correct activation and the centralization of the TLS; **B,** TLS shape when positioned symmetrically in an interbracket distance of 23 mm and activated 5 mm.

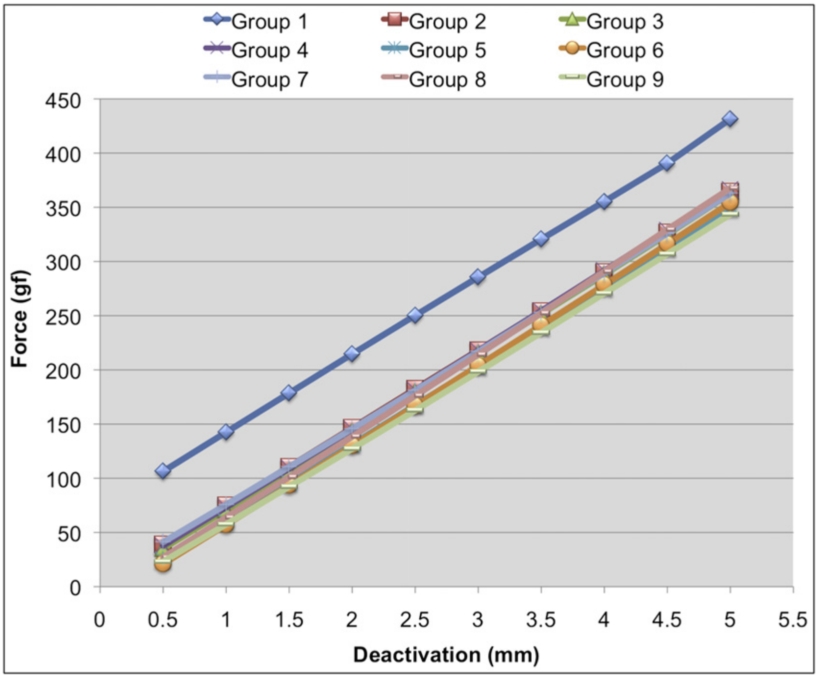

Fig 6. Chart depicting the average forces produced during deactivations from 5 to 0.5 mm for the groups tested.

Table I. Multivariate profiles test significances for the force, moment, and MF variables

Variation	*Force* P *value*	*Moment* P *value*	*MF* P *value*
Time	<0.001	<0.001	0.600
Deactivation	<0.001	<0.001	0.010
Deactivation × time	0.106	<0.001	0.464

TLSs in "neutral position." The post-hoc Tukey test, at a level of 5%, was used to identify the groups' differences.

RESULTS

There was a significant decrease of force from time among the groups when the total profiles of the TLSs were compared (P <0.001). The horizontal forces were higher for group 1 compared with the other 8 groups (Table I, Fig 6), which were not different among themselves. Group 1 showed a total profile average force of 267.6 gf compared with 200.7, 197.8, 200.5, 186.1, 186.2, 200.2, 196.1, and 181.7 gf, from groups 2 through 9, respectively (Table II). There was no significant interaction of time on the rate of force decrease of the loops (P = 0.106) among the groups (Table I).

The amount of overlap of the vertical extensions of the TLSs ("neutral position" in Table III) was greater in group 1 (−0.99 mm) than in the other groups (P <0.001). Time did not have an effect on the LD ratio of the TLSs, which were the same in all groups (P = 0.159) (Table III), ranging from 70.8 to 75.5 gf per .5 mm.

The total profile average moment levels produced throughout the deactivation were higher for group 1 (1932.6 gf.mm) than for the other groups (P = 0.001). The moments recorded in groups 2 through 9 were similar among themselves, ranging from 1471.3 (group 3) to 1636.4 gf.mm (group 6). There was a significant interaction between evaluation time and deactivation (P <0.001) (Tables I and II, Fig 7).

All groups produced similar total profile average MF ratios (P = 0.600); the averages from groups 1 through 9 were, respectively, 8.3, 11.2, 26.7, 12.2, 14.2, 10.2, 14.7, 18.3, and 18.2 mm. No interaction was found between time of evaluation and deactivation (P = 0.464) (Tables I and II, Fig 8).

DISCUSSION

Group 1 showed higher horizontal forces than did the other groups. After the initial reduction in the first 24 hours, the behavior of all other groups was similar, with little influence of time on the deactivation forces produced. This can be explained by the stress-relaxation phenomenon. The greater decrease of force in 24 hours compared with measurements taken over longer periods of time agrees with several other reports that measured this effect, showing that it is time-dependent.[9,14,16]

Table II. General profiles means and standard deviations for the forces and moments

	Force		*Moment*		*MF*	
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*
Group 1	267.6 A	105.2	1932.6 A	259.2	8.3	3.2
Group 2	200.7 B	111.5	1537.9 B	249.9	11.2	18.2
Group 3	197.8 B	107.8	1471.3 B	239.3	26.7	166.7
Group 4	200.5 B	111.1	1616.5 B	250.0	12.2	34.5
Group 5	186.1 B	104.8	1563.1 B	206.7	14.2	13.7
Group 6	186.2 B	110.4	1636.4 B	236.3	10.2	33.4
Group 7	200.2 B	108.4	1511.6 B	269.5	14.7	47.1
Group 8	196.1 B	112.5	1582.7 B	234.5	18.3	32.3
Group 9	181.7 B	109.5	1553.9 B	349.9	18.2	45.1

Different letters indicate group differences.

Table III. Means and standard deviations for overlapping of the vertical extensions of the TLSs and LD ratios

	Neutral position (mm)*		*LD† (gf/0.5 mm)‡*	
Time	*Mean*	*SD*	*Mean*	*SD*
Group 1	−0.99 A	0.31	71.7	3.2
Group 2	−0.03 B	0.55	71.9	2.3
Group 3	0.01 B	0.36	72.4	2.6
Group 4	−0.01 B	0.45	72.6	4.5
Group 5	0.16 B	0.02	72.0	2.7
Group 6	0.22 B	0.37	73.7	3.9
Group 7	−0.08 B	0.57	71.1	4.0
Group 8	0.15 B	0.32	75.5	3.6
Group 9	0.20 B	0.46	70.8	4.7

*$P < 0.001$; †$P = 0.159$; ‡To acquire the LD per millimeter, multiply the values by 2.

However, this effect had not yet been shown on wires in more elaborate configurations such as loops. Nevertheless, clinicians should expect a drop of about 15.5% in the force levels of TLSs in the first 24 hours after placement (Table IV). This study shows that the force system currently proposed in the literature can be reviewed since, in only 24 hours, the values of the horizontal forces and moments are reduced, on average, by 15.5% and 17.15%, respectively (Table IV).[7,10]

The behavior of overactivation caused by the "neutral position" (due to the overlap of the vertical extensions of the TLSs), which decreased over time, was consistent with the horizontal forces' decrease and can be used to explain it. Although there are no reports of a similar evaluation over time in the literature, this was an expected effect, because the opposite situation (increase of force) is expected with an increase on any loops' preactivation because of an increase in the vertical extensions' overlap.[17] Normally, an opening of the internal "ears" of the loop should be used to compensate for it.[7] Nevertheless, this effect should be considered, because nonoverlapping TLSs could experience underactivations by 24 hours after their engagement on the patient's brackets.

The LD ratio was constant throughout the evaluation periods and did not seem to be affected by the deformation of the TLSs. This was demonstrated by both the lack of interaction between time and deactivation (force variation or LD ratio) (Table I) and the undetected differences when the LD ratios were compared among the TLSs (Table III). Deformation on the geometry of the structure of a spring can have different effects on the force systems, as demonstrated when different cantilever setups were evaluated.[18] Depending on the deformation of a TLS caused by preactivation, it might assume a slightly different LD ratio when the TLS is tied to the brackets. In our study, the difference in shape was probably too small to cause a significant effect.

There was also a significant decrease in the moments produced by the TLSs over time. It is known that an increase in the angle (preactivation intensity) of the horizontal extremities of a loop causes an increase in the moment produced by it.[17,19-22] Thus, in 24 hours, there was deformation of the spring's horizontal extremities, causing the angle formed by these extremities to be reduced and lower moments produced. So, as already mentioned, clinicians should be aware that, when

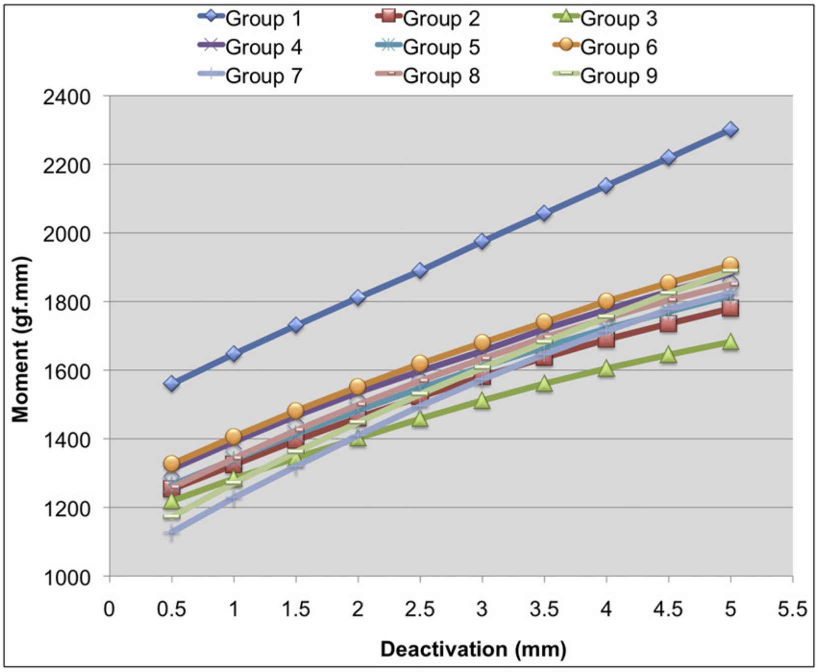

Fig 7. Chart depicting the average moments produced during deactivation from 5 to 0.5 mm for the groups tested.

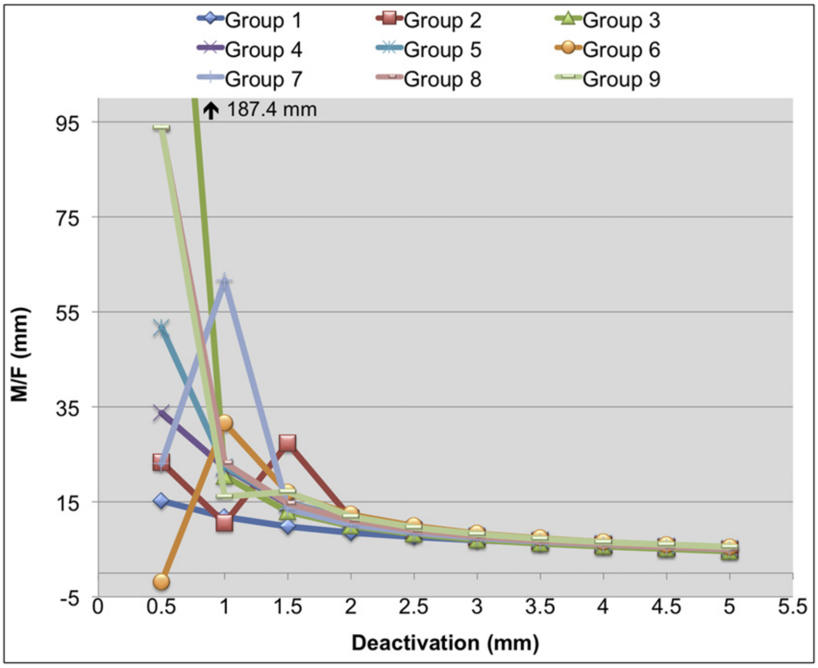

Fig 8. Chart depicting the average MF ratios produced during deactivation from 5 to 0.5 mm for the groups tested.

Table IV. Means and standard deviations for the forces, moments, and MF ratios at deactivation from groups 1 and 2

	Distance (mm)	Group 1 (immediate)		Group 2 (24 hours)		
		Mean	SD	Mean	SD	Difference
Force	5.0	431.5	23.1	364.4	44.6	67.1
	4.5	390.6	21.3	326.9	44.1	63.7
	4.0	355.4	21.0	290.3	43.6	65.1
	3.5	320.6	20.9	254.1	43.0	66.5
	3.0	285.5	20.7	218.1	42.5	67.4
	2.5	250.3	20.6	182.4	42.0	67.9
	2.0	214.7	20.8	146.5	41.2	68.2
	1.5	178.6	21.0	110.6	40.3	68.0
	1.0	142.5	21.5	74.8	39.4	67.7
	0.5	106.4	22.0	38.9	38.5	67.5
						Average 66.9 (15.5%)
Moment	5.0	2301.2	124.4	1779.7	190.8	521.5
	4.5	2218.3	122.1	1735.5	186.0	482.8
	4.0	2137.8	113.8	1689.4	183.6	448.4
	3.5	2056.5	108.2	1636.5	183.0	420.0
	3.0	1974.9	104.5	1581.2	184.7	393.7
	2.5	1889.4	104.3	1522.2	185.0	367.2
	2.0	1811.0	102.4	1461.7	191.3	349.3
	1.5	1730.0	104.0	1395.4	198.5	334.6
	1.0	1646.8	107.7	1324.5	205.8	322.3
	0.5	1559.8	113.0	1252.8	212.5	307.0
						Average 394.7 (17.15%)
MF*	5.0	5.3	0.4	4.9	0.7	
	4.5	5.7	0.4	5.4	0.8	
	4.0	6.0	0.5	5.9	1.0	
	3.5	6.4	0.5	6.6	1.3	
	3.0	6.9	0.6	7.5	1.9	
	2.5	7.6	0.7	8.9	3.1	
	2.0	8.5	0.9	11.5	6.7	
	1.5	9.8	1.2	27.3	49.3	
	1.0	11.8	1.8	10.5	17.7	
	0.5	15.2	3.1	23.3	15.2	

*No differences were found among the groups.

changes happen in the preactivation of the TLS, this might have an effect on the horizontal force, since we are reducing the overlap of vertical extensions and consequently the activation extension; therefore, this effect should be compensated for.[23]

Unlike the horizontal force, there was a significant interaction of time on the variation of the moments with deactivation (P <0.001). The deactivation patterns for all groups were not similar; ie, the amount of moment decrease for every 0.5 mm of deactivation was different with time. Figure 5 shows that the moment levels during deactivation fit initially on a relatively straight line, which, over time, became a polynomial curvature showing less decrease. This is an interesting effect that has not been shown in the literature and is explained by localized deformation in a specific part of the loop, which affects moments and forces differentially.

The MF ratios remained unchanged in the periods studied, and there was no interaction of time on the increase rate of the MF ratios levels with deactivation. This maintenance was probably because the stress relaxation in the wire happened only on the horizontal extremities of the TLS, causing a decrease of the moment and on the overlapping of the vertical extensions of the TLSs, which proportionally decreased the force levels.

The MF ratios became inconsistent and unpredictable in the final stages of the deactivation of the TLSs, from 1 to 0 mm (Fig 8). This was due to the stress relaxation of the wire, which caused underactivations of the TLSs of all groups, except group 1. This means that, in the "neutral position," the vertical extensions of some TLSs might not be touching. Since the testing machine tested the loop to 0 mm of extension, some TLSs registered a negative force, which would not occur clinically, since these loops would have been reactivated long before that. Therefore, this effect, seen in Figure 6, would not occur; instead, the MF ratios would have become higher, not because of negative values but because of low force levels, maybe even causing uncontrolled tipping, with the crown of the tooth moving in the opposite direction of the force. Clinically, from all the discussed data, this would mean that, when using TLSs, the inner "ears" should be opened less than normally indicated in the literature, generating an extra millimeter of overlap of the vertical extensions than the amount of overlap desired, compensating for the effect caused by the stress relaxation of the TLSs, as demonstrated in this study.

CONCLUSIONS

The TLSs preactivated by concentrated bends suffered progressive deformation over time. This effect was critical on the first 24 hours on (1) the moment reduction, (2) the decrease in the rate of moment reduction, and (3) the decrease in the overlap of the vertical extension of approximately 1 mm, causing a horizontal force reduction at a given activation.

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