

Laboratorial study of the cuspid's retraction timing and tipping effects during space closure, using the segmented arch technique

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Abstract

Objective: Evaluate the cuspid's retraction time and tipping effects, after submitting it to three different orthodontic retraction loops: the "T" loop, the "boot" loop, and the "tear drop" loop. **Methods:** It was used the following orthodontic wires: Morelli 0.019 x 0.025-in stainless steel, 3M Unitek 0.019 x 0.025-in stainless steel and Ormco 0.019 x 0.025-in beta-titanium (TMA™). The resulting sample from the combination of these variables was submitted to a test developed on a typodont simulator used specifically for this purpose. **Results:** As the closure timing concerns, it was verified that a slower closure and therefore, a smaller releasing force system was achieved by the "T" loop design and by employing the beta-titanium alloy on its construction. As to the tipping effects generated by the retraction device, the "tear drop" loop caused greater tipping effects than the other loops evaluated. The "T" loop, on the other hand, showed itself statistically related to the lowest tipping numerical values. However, when the 3M Unitek stainless steel wire was used to produce the device, all of the types of loops evaluated were considered statistically similar. **Conclusion:** Regardless of the loop design, the ones built out of beta-titanium alloy kept them statistically related to the lowest tipping numerical values observed for the retracted dental element.

Keywords: Orthodontics. Segmented arch. Orthodontic space closure.

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INTRODUCTION AND LITERATURE REVIEW

During the orthodontic treatment it is expected that an optimal force used to promote dental movement provides a satisfactory result in a reasonable period of time, with minimum damage to the adjacent structures and minimal discomfort to the patient.^{1,10,17,21,27,29} It seems to exist a wide range of force values that produces a maximum amount of movement of the dental element,²⁷ without undesirable movement of the anchorage unit.^{10,12,13,21}

Several devices can be used to obtain dental movement.^{2,3,6,12,18,22-25,27,28} One can choose between a sliding mechanics along continuous orthodontic arches and frictionless mechanics, where segmented arches with orthodontic loops can be used.^{10,28} However, in both cases, it is not possible to eliminate the rotational and tipping components from teeth, due to the fact that the accessories of the orthodontic device are positioned some millimeters labial to the tooth axis and a few millimeters occlusal to the center of resistance of the teeth.¹⁰

Some physical concepts need to be revised so that one can understand the relationship between the forces and the dental movement.^{14,15,20,26} Each object or body has a point where it can be balanced perfectly, which is known as center of gravity of the object. However, the teeth have an additional complication. They are surrounded by periodontal structures that involve the root, but not the crown. Then, another point has been used: the center of resistance. It is important to point out that the position of the center of resistance varies with the root length and the alveolar bone height as well.^{20,26} Generally, the tooth can move in three ways: translation or body movement; pure rotation movement, where the tooth will rotate around its center of resistance, and combined translation-rotation movement.^{13,14,20,26}

The authors defined the moment of force as the magnitude of force multiplied by the perpendicular distance to the action line of that force

to the center of resistance of the tooth.^{14,15,20,26} If the line of action of an applied force does not pass through the center of resistance of the dental element, the force will produce some rotation on that tooth. This rotation potential is called moment. The orthodontist creates a binary of forces in the bonded device, which will oppose to the moment produced by the force acting on the dental element, so that the forces act directly on the center of resistance of the tooth.²⁶ The dental movement is determined by the ratio between the binary moment (M) used to control the position of the root and the force (F) used on the crown to move the tooth. The more heavy is the force, greater is the moment of the binary (on the accessory) needed to maintain the desired rotation.²⁰ In a M/F ratio of 5:1, an uncontrolled tipping occurs; with a M/F ratio of 8:1, a controlled tipping occurs; in a M/F ratio of 10:1, translation occurs; in a M/F ratio of 12:1, root movement occurs.^{8,13,20,26,28}

Several authors discussed the optimal properties of devices used for dental movement.^{2,8,9,17} Among the respective properties are:

1. It should generate appropriate levels of force, a low load/deflection ratio^{2,16,20,23,25} and a high M/F ratio, in order to reach the desired dental movement. Gable or antitipping bends can be incorporated to the devices in order to increase the level of moment produced.^{7,8,17,24,28} This reflects an increase of the M/F ratio; differential moments can still be generated, changing the positioning of the devices.^{3,12,28}

2. It should be capable to submit to a reasonable range of activation/deactivation, liberating relatively continuous forces and moments.

3. It should be sufficiently small to adapt comfortably in the available intra-oral space.

In addition, the properties of the devices can be changed with modifications in the thickness, shape, amount of wire used, and rate of activation according to the modulus of elasticity of the wire by thermal treatment.¹⁷

In order to reach these objectives, several devices with different configuration have been introduced in the literature.^{3,7,12,16,18,23,24,25,28,30} The results provided by these different devices can be linked to relevant factors of the orthodontic treatment, i.e., the time needed for accomplishment of the dental movement, as well as tipping effect on the dental element. Another factor to be taken in consideration is the type of wire used to build the orthodontic device. There are several types of wire available in the market,^{3,8,9,11,12,16,18,19,29,30} which possess different features and mechanical properties.

The current work aimed to evaluate the retraction rate and the degree of tipping suffered

by the moved dental element using three different types of orthodontic retraction springs: the “T” loop, the “L” loop, and the “tear-drop” loop. For making these springs different materials were used: two commercially available stainless-steel wires, and one commercially available beta-titanium wire (TMA™).

MATERIAL AND METHODS

In the present study, three types of loops, the “T” loop, the “L” loop, and the “tear-drop” loop, conformed in stainless steel wires (Morelli, Sorocaba, SP, Brazil and 3M Unitek, Saint Paul, MN, USA) and one beta-titanium wire (TMA™, Ormco, Orange, CA, USA) were evaluated.

TABLE 1 - Description of the sample groups.

GROUP	DRAWING OF THE LOOP	WIRE TYPE	COMMERCIAL MARK
A	“T” loop	Stainless Steel	Morelli
B	Tear-drop	Stainless Steel	Morelli
C	“L”	Stainless Steel	Morelli
D	“T” loop	Beta-titanium (TMA™)	Ormco
E	Tear-drop	Beta-titanium (TMA™)	Ormco
F	“L”	Beta-titanium (TMA™)	Ormco
G	“T” loop	Stainless Steel	3M Unitek
H	Tear-drop	Stainless Steel	3M Unitek
I	“L”	Stainless Steel	3M Unitek



FIGURE 1 - Typodont model used for evaluation: simulation of exodontia of 44 (set-up of right lower arch).



FIGURE 2 - Anchorage unit stabilized with gypsum (coated with clear nail varnish).



FIGURE 3 - Metallic teeth with the respective accessories positioned.

A partial typodont assembly simulating a right lower arch was made for the experiment. The created model simulated the exodontia of the element 44 (Fig 1). The elements 47, 46 and 45 were fixed with dental gypsum and represented the posterior anchorage (Fig 2). Teeth # 47, 46, 45, and 43 received Edgewise standard 0.022 x 0.030-in slot brackets. Element 43 received a vertical segment of wire, welded orthogonally to the slot to serve as a reference for reading tipping suffered by this tooth during the proposed movement (Fig 3).

Three types of loop were conformed to each wire (all wires were 0.019 x 0.025-in), yielding nine different evaluation groups (Table 1). To help in the making of the arch segments with loops, it was used a chart where was drawn the outline of the arch segment and a template with the drawing of the loops (Figs 4 and 5). Fifteen samples were

made for each group (135 arch segments evaluated) (Fig 6).

Then, each of the arch segments was tested, according to the following sequence:

1. The arch segment was tied to the assembly with 0.010-in stainless steel ties (Fig 7).

2. This condition was recorded with photographs (T1). The mannequin was stabilized in this moment by means of a support with screws. Then, the distance between the anterior border of the support to the most anterior portion of the photo camera lens was standardized: i.e. 12.4 mm, so that both elements were parallel to each other, from a upper view (Fig 8). The lens opening was adjusted to "32" and the shutter speed was "90". Two gridlines demarcated in the base of the articulator were used to standardize the framing of the photographs.

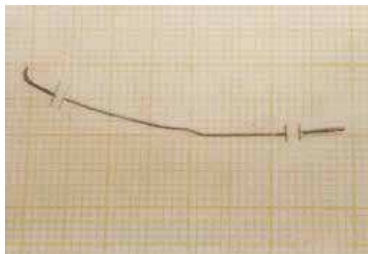


FIGURE 4 - Chart used to make the archwires segments.

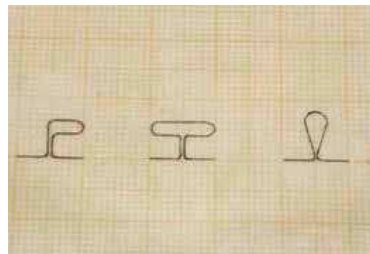


FIGURE 5 - Template used to make the loops.

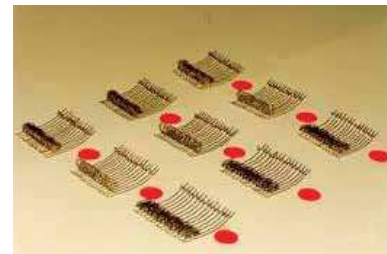


FIGURE 6 - Sample used in this work (n = 135 archwires).



FIGURE 7 - Archwire being fixed to the assembly with steel tie.



FIGURE 8 - View of the assembly in place for photographic recording.



FIGURE 9 - Activation of the loop with a tieback with double tie in the omega loop. The amount of the activation was controlled with a divider.

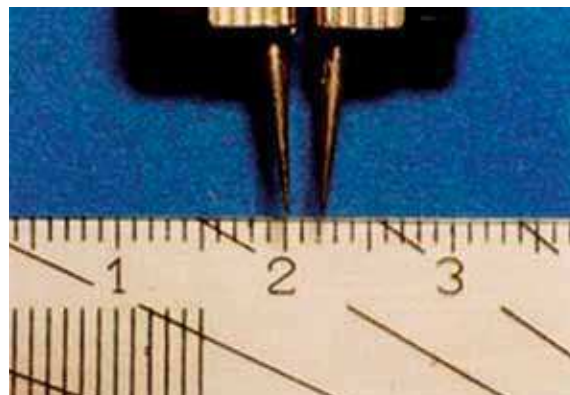


FIGURE 10 - Measurement of the amount of activation with divider and scale: 2 mm of activation.

3. The spring was activated by means of a tie-back, promoting the movement of the distal segment to that direction. The spring was opened until an opening of 2 mm was achieved, checked by means of a divider (Figs 9 and 10).

4. The articulator was then immersed in a recipient with warm water (50°C), in order to allow the deactivation of the spring.

5. Immediately after the immersion, the chronometer was started. The time (in seconds) required for complete deactivation of the spring was recorded by visual inspection.

6. The articulator was positioned again in the support so that a new photo was obtained (T2).

7. It was necessary that the cuspid has assumed repeatedly the same initial position to the procedure could be considered reproducible for each one of the arch segments. This was achieved with a segment of 0.0215 x 0.0275-in ideal arch, used as a guide for repositioning of the cuspid after the evaluation of each arch.

8. The assembly was immersed in cold water to evaluate a new arch.

In order to avoid that possible alteration of the characteristics of the wax after successive evaluations could interfere in the fidelity of the results, the evaluation was accomplished in the following manner: the nine combinations were divided in 3

TABLE 2 - Sequence of evaluation of the devices.

	1 ST TYPE OF LOOP EVALUATED	2 ND TYPE OF LOOP EVALUATED	3 RD TYPE OF LOOP EVALUATED
Morelli wire loops (new wax)	"T"	"L"	"Tear-drop"
Ormco wire loops (1st change of wax)	"L"	"Tear-drop"	"T"
3M Unitek wire loops (2nd change of wax)	"Tear-drop"	"T"	"L"

groups, separated by the type of wire. The wax was replaced for each type of wire. In addition, the arrangement of type of the loop to be evaluated was changed, according to sequence described in Table 2.

Having the photographic recordings of the initial (T1) and the final (T2) conditions of the assembly, a tracing paper was placed over these pictures. The long axis of the cuspid was traced, and the line was extended until contact the grid-line demarcated in the base of the articulator,

from which the upper margin was traced. Then, the angle formed between these two lines was measured for all arches evaluated, for both the initial (T1) and the final (T2) conditions. The difference between these two values could be calculated, and the angular variation presented by the cuspid with the closure of the loop was obtained. Another variant recorded was the time required for the deactivation of the loop.

The results were recorded in individual forms for data collection, and ultimately submitted to statistical analysis using non-parametric comparison tests (Kruskal-Wallis). The conditions tested were the loop design (independent of the type of wire) for the three groups, the type of wire (independent of the loop design) for the three groups, and finally the interaction between the loop design and the type of wire for the nine groups.

RESULTS AND DISCUSSION

To help the analysis and discussion of the results, two issues were addressed, according with two evaluated variables: the time for closure of the loop and the degree of tipping suffered by the tooth. Inside each topic, it was evaluated the effects of the different types of wires (independent of the drawing of the loop), the effects of the different loop types (independent of the wire type) and the effects of the interaction between loop type and wire type, on the variant in question. When interaction was verified, post analysis was performed, to investigate whether the effect occurred due to the loop, to the wire, or both.

Time of closing of the loop

The time observed for closing of the loop was recorded in seconds. In an attempt to guide the discussion of this topic on which device exerted a larger or minor force on the tooth to be moved, it was taken into account that less time for closing of the loop is related to a higher force released by the loop. Conversely, the smaller the force, the more is the time required for closing of the loop.

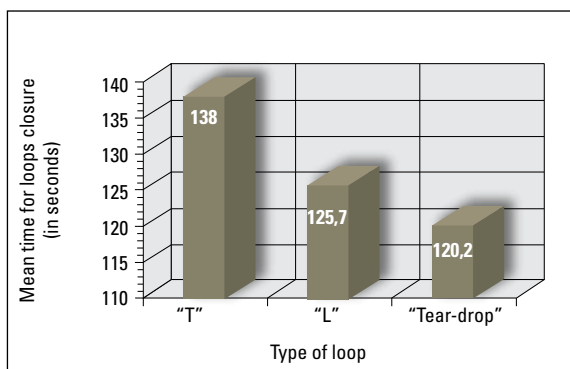
Burstone³ reported that the optimal force for dental movement is that capable to produce a fast movement with minimal discomfort and damage to the tissues, using continuous and slight forces. Hixon et al¹⁰ stated that the fast tooth movement generated when using light forces seems to be a result of tipping movement that produces great pressure on the alveolar crest.

All recordings obtained in this work were subjected to statistical analysis. Two segments presented values for this variable that were characterized as outliers from all appraised arch segments. These values were excluded from the analysis. (Specimen # 14, Group H; Specimen # 1, Group F). The analysis was subsequently performed for the interaction between loop type and wire type for the variant "time of closing of the loop". The interaction was not statistically significant. On the other hand, the variables wire type and loop type were significantly different when analyzed independently:

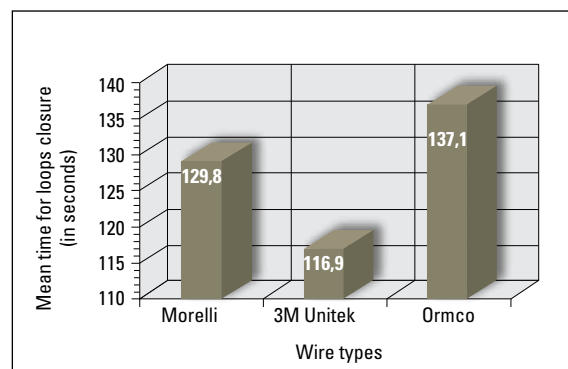
Relationship of the loop type with the variant "time of closing of the loop"

According with the values of Graph 1, it was observed that the "T" loop took more time to accomplish the tooth movement, therefore exerting a smaller force on the cuspid than the other loops. In spite of the "L" loop exerted less force than the "tear-drop" loop, the difference was not statistically significant.

The good performance of the "T" loop was previously reported by Burstone and Koenig.⁵ They stated that this loop uses a great amount of wire for its construction, especially cervically. This loop configuration with great amount of wire arranged horizontally at the cervical, even when is built with stainless steel wire, yields a significant decrease of the load/deflection ratio. This was also observed more recently by Shimizu et al.²³ The authors concluded that the "T" loop is capable to generate relatively low load/deflexion rates, and more consistent magnitudes of force during the



GRAPH 1 - Mean time (in seconds, y-axis) for deactivation of loops with different drawings (x-axis).



GRAPH 2 - Mean time (in seconds, y-axis) for deactivation of different commercially available loops (Morelli and 3M Unitek: stainless steel; Ormco: beta-titanium; x-axis).

deactivation as a result. Souza et al²⁵ also supported the use of "T" loops by orthodontists; as well as several authors did previously.^{3,28}

Relationship of the wire type with the variable "time of closing of the loop"

In 1979, Burstone and Goldberg⁹ introduced to the market a beta-titanium alloy, considered as the newest material in the orthodontic profession.

Since then, this alloy became an option for the orthodontist, with characteristics that surpassed other alloys, such as: capacity of application of light forces, continuous deactivation of the force with time, higher precision in the application of a force and the capacity of application of larger activations, associated to an extended "working time" of the device. In 1980, the importance of this alloy was advocated due to its great potential in orthodontics.⁴ The main reason is that in an orthodontic device, the maximum elastic flexion increases with the accumulated force/modulus of elasticity ratio of the material. Beta-titanium alloys possess one of the highest values for this ratio (about 1.8

times higher than that of stainless steel), while maintains good formability.

The importance ascribed to beta-titanium alloys by these authors was confirmed in the present work. According with the values of Graph 2, the stainless steel devices accomplished the dental movement more quickly than beta-titanium alloy devices. This was reported earlier by Stagers and Germane,²⁸ that found that the load/deflexion ratio can be changed by differences in wire composition. TMATM loops have low modulus of elasticity, and a lower load/deflexion ratio than stainless steel loops. This was also reported earlier by Boshart et al² that found that there was a change in the rigidity of coil springs with different compositions. Menghi, Planert and Melsen¹⁸ also compared systems of force liberated by beta-titanium and stainless steel devices, and found a conclusion similar to our study: beta-titanium devices released 40% of the force provided by identical stainless steel loops. Beta-titanium alloy loops are preferable in comparison to stainless steel loops due to their higher activation range and consistent

liberation of forces.

It could be concluded from our results that the beta-titanium alloy devices exerted less force on the cuspid than the others. This is a clinical information that is extremely important. Manhartsberger, Morton and Burstone¹⁶ reported that orthodontic extraction therapy is common in adult patients. In these cases, where bone loss is a complication, they suggest the use of beta-titanium alloys for loops. These alloys reduce the magnitude of the forces applied to the teeth and yield a lower load/deflection ratio (allowing the making of an arch with smaller rigidity). Another factor to be considered with the use of less rigid wires, according to the authors, is the potential of increasing the amount of activation of the loop. Burstone³ stressed out that a beta-titanium alloy for making loops for closing of spaces is more easy to handle, allows for a simplification in the drawing of the loop and has a low load/deflection ratio. This means that it can release optimal levels of force, which are dissipated slowly, with great amounts of activations. The clinical relevance of this issue is that with great activations, an error of 1 mm during activation is not as significant as the same error in a more rigid device.

The lack of statistical relationship between stainless steel arches remains unclear. Further investigation is required about the proportions of alloy's components in manufacturing process. In spite of this limitation, beta-titanium alloy arches exerted less force on the tooth, supporting the findings of Kapila and Sachdeva.¹¹ Also in this work, commercially available beta-titanium wires, known as TMA™, presented lower elasticity modulus than stainless steel and chromium-cobalt wires, and approximately the double of the presented by nickel-titanium wires. Therefore, beta-titanium alloy wires can be deflected without permanent deformation (about two times) than stainless steel wires, have higher formability than nickel-titanium wires, and allow that loops can be incorporated to the wire. According to the same authors, its only

disadvantage is the high level of friction presented when it is in contact with the bracket.

However, it is proper to stress out here that a low load/deflection ratio is not necessarily advantageous for the dental movement in all stages of the orthodontic treatment. According to Yang, Kim and Kim,³⁰ low rigidity nickel-titanium wires are recommended in the early stages of the treatment, beta-titanium wires are recommended in the intermediary stages due to their moderate rigidity, and high-rigidity arches are more appropriate for the final stages. Therefore, the relevance of alloys with high load/deflection ratio cannot be omitted by the results presented by our study, as showed by the stainless steel wires throughout the orthodontic therapy.

Degree of tipping of the cuspid

The variation in tipping of the cuspid after its movement can be attributed to the fact that the point of application of forces (bracket) is placed far from the center of resistance of the element in a cervico-occlusal direction. This generates a moment in the tooth to be moved, tipping it.

Despite it is not the objective of the present discussion, it is also convenient to highlight that the point of application of forces of the evaluated devices on the tooth is far from its center of resistance in the labial-lingual direction, which is responsible for the tendency of rotation of the tooth during the movement.

Hixon et al¹⁰ reported the difficulty in eliminating the rotational and tipping components presented by the element to be retracted, due to the distance of the center of resistance to the point of force application. In our work, the control of the variation in the angulation of the cuspid could be attributed to the M/F ratio of the devices used. According to Smith and Burstone,²⁶ the aim is to create a binary of forces in the accessory bonded to the tooth, opposing the moment produced by the force that acts on the tooth. The type of movement of a tooth is determined by the ratio between the

magnitude of the binary (M) and the force (F) applied on the bracket. Kuhlberg and Priebe¹³ mentioned that a small value in this ratio (about 7/1) provides a movement of controlled tipping; a ratio of approximately 10/1 is capable to promote translation of the tooth. Higher values (about 12/1) can cause movement of the root apex while the crown of this element remains stable.

Regarding the issue “degree of tipping of the cuspid”, the presence of an outlier was verified (Specimen # 15, Group F). This value was eliminated to not compromise the final result of the analysis. The analysis revealed that the interaction between the loop type and the wire type with the variable “degree of tipping of the cuspid” was statistically significant. The unfolding of the results was the following:

Relationship of the loop type with the variable “degree of tipping of the cuspid”

In function of the post analysis, the loop types were evaluated for each type of wire, separately, according to the Graph 3.

Staggers and Germane²⁸ stated that the drawing of the retraction spring influences the load/deflection ratio. In a general way, according to the Graph 3, the teardrop loop promoted greater tipping of the tooth than the other evaluated loops. On the other hand, the “T” loops presented, within each wire type, statistical correlation to the smallest variation values in angulation presented by the cuspid after its retraction. The exception was the third evaluated type of wire that did not presented statistically significant differences.

The variation in the angulation of the cuspid can be attributed to the M/F ratio of the devices: a higher magnitude force provided by the teardrop loop yields a low M/F ratio. Then, one could expect that “T” loops are capable to yield smaller magnitudes of force and, consequently, provide a higher M/F ratio, tipping less the tooth to be moved.

Burstone and Koenig⁵ suggested that to in-

crease the M/F ratio of a loop during activation the length of the loop in an apical direction should be raised. Another manner is to increase the amount of wire used in the terminal segment of the loop, which decreases the load/deflection ratio. According with the same authors, this can be achieved by using the “T” loop. However, according with our results, even a “T” loop seemed to be unable to avoid tipping of the cuspid during the movement. This undesirable effect can be minimized by the incorporation of compensatory folds in these loops (gable bends), in order to promote a greater root movement.

Manhartsberger, Morton and Burstone¹⁶ reported that introducing angulations in the loop could increase the M/F ratio of a device.

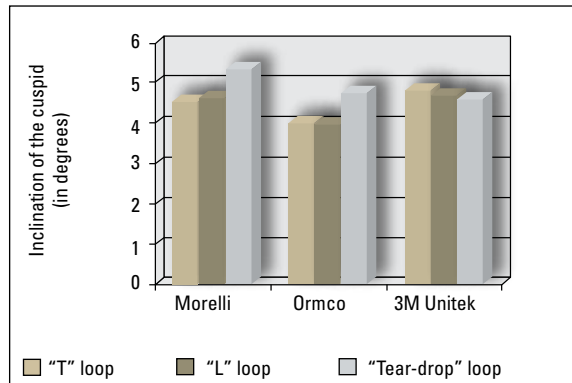
Staggers and Germane²⁸ showed that, even for a “T” loop, it is very difficult to get a 10:1 M/F ratio, required to obtain translation movement, without making gable bends. The incorporation of this type of bends was also suggested by Burstone,³ Chen, Markham and Katona,⁷ Faulkner et al,⁸ Shimizu et al,²³ and Souza et al,²⁵ among others.

It would be interesting, in a next study, to use a method similar to the present work including gable bends in the tested loop, for evaluation of the advantages from these bends.

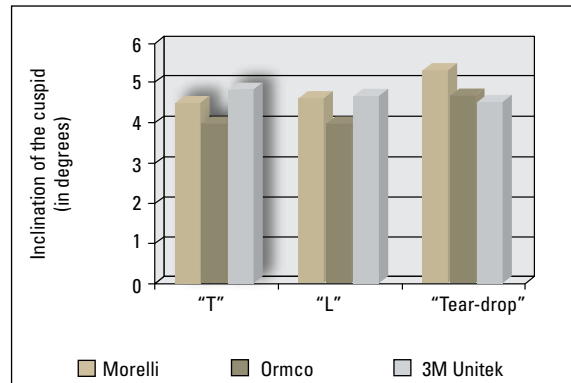
Relationship of the type of wire with the variable “degree of tipping of the cuspid”

Investigating the influence of different resources available to the orthodontist for obtaining a device that could yield a suitable M/F ratio, it was aimed to evaluate the influence of the alloy’s type used for fabricating the loop on the tipping presented by the cuspid after retraction. Regarding this topic, due to the unfolding of the results, the wires of different composition were evaluated separately for each type of loop, according to the Graph 4.

It was observed that, in a general way, the beta-titanium alloy loops were statistically related, in all the 3 groups, to the smallest values of variation



GRAPH 3 - Inclination of the cuspid after retraction (in degrees, y-axis) for different drawings, according to each type of wire (x-axis).



GRAPH 4 - Inclination of the cuspid after retraction (in degrees, y-axis) for different commercially available archwires (Ormco: beta-titanium; 3M Unitek and Morelli: stainless steel), for each loop drawing (x-axis).

in angulation presented by the cuspid after retraction, when compared to the stainless steel loops evaluated in this study.

In spite of Stagers and Germane²⁸ have affirmed that the M/F ratio is not influenced by the composition of the wire used, it could be expected that the use of more resilient wires provides smaller magnitudes of force. According to Shimizu et al,²³ devices capable to generate relatively low load/deflection ratios, provide more consistent magnitudes of force during deactivation as a result; yielding high moment/force (M/F) ratios, and ultimately more root movement.

The combination of materials with lower modulus of elasticity and rigidity, associated to a loop drawing capable to decrease the load/deflection ratio of the assembly, can produce devices that promote a slower closing of the loop after its activation and a retraction with lighter, biologically more compatible forces. Thus, smaller magnitudes of force can act in the M/F ratio increasing its values and, consequently, decreasing tipping effects generated by the forces of dental movement. These forces do not act directly on the center of

resistance of the teeth submitted to the orthodontic treatment.

However, this study showed that even the combination of a loop drawing capable to promote a lower load/deflection ratio with more resilient wires was unable to isolate tipping effect suffered by the moved tooth. Nevertheless, it is likely that additional resources should be used seeking this objective, such as the incorporation of bending in these devices.

Another important issue to be considered for discussion is that the high tipping values recorded at the end of the retraction procedure maybe are due to the fact that it has not awaited sufficient time so that the evaluated devices could release all its potential of root movement. Stagers and Germane²⁸ reported that since the M/F ratio increases as the loop is deactivated, the loop should not be reactivated so frequently. According to the authors, repeated reactivations do not allow that the loops reach a sufficiently high M/F ratio to promote a translation movement of the tooth. It would of interest that this fact was taken in consideration in the case of further investigation.

CONCLUSION

According to the results obtained in this work, it could be concluded that:

Time of closing of the loops

There was no interaction between the type of wire and loops for this variable. However, when considered independently, the differences were significant:

- Loop type: the "T" loop take more time to deactivate than the others.
- Wire type: the beta-titanium alloy loop takes more time to deactivate than the others.

Degree of tipping of the cuspid

In this case, it was observed an interaction between the type of the loop and wire. The post analysis revealed was accomplished as following:

Loop type

The "teardrop" loops promoted greater dental tipping than the others evaluated. On the other hand, the "T" loops showed statistical correlation to the smallest tipping values. However,

when 3M Unitek stainless steel wires were used to make the loops, the 3 types did not present statistical difference for this variant.

Wire type

The beta-titanium alloy loops were statistically correlated to the smallest tipping values observed for the moved tooth, regardless of the loop drawing used.

Therefore, the combination of a material with lower modulus of elasticity and rigidity (beta-titanium) associated to a loop drawing that uses greater amount of wire (such as "T" loops) produces a device that generates a relatively lower load/deflection ratio. As a consequence, this provides lighter and consistent force magnitudes during deactivation, increasing the moment/force ratio, providing greater root movement.

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