Review Article

Closing Loops; Geometry and Mechanical Properties: A Review

Marcelo do Amaral Ferreira,[1] Fábio Rodrigo Mandello Rodrigues,[2] Marco Antônio Luersen,[3] Paulo César Borges[4]

[1]DDS, MSc, PhD. Private Practice.

[2]Professor Dr. Eng.- Academic Department of Mechanical Engineering.
DAMEC-UTFPR (Technical Federal University of Paraná- Campus Pato Branco), Brazil.

[3,4]Professor Dr. Eng.- Academic Department of Mechanical Engineering
DAMEC-UTFPR (Technical Federal University of Paraná- Campus Curitiba), Brazil.

Corresponding author: Marcelo do Amaral Ferreira; Email: regunteriat@gmail.com

ABSTRACT

Introduction: To verify the literature concerning the mechanical properties, measuring methods and the resultant force system developed by closing loops (CL). The effectiveness of a certain CL is related to its geometry and the nature of the wire material. To obtain the necessary performance CL must work in the elastic range and be geometrically configured to express an adequate force system, which is the result of compensatory gable bends and preactivation. It was performed a literature search in MEDLINE from 1974 to 2014 using search for "orthodontic AND retraction AND springs" and "orthodontic AND closing AND loops". From the 147 papers resulted from Pubmed, 50 were chosen. Overall, many works seek not only improve the knowledge about the force system, but also a better understanding of their mechanical behavior. FEM, Holographic and Photoelastic studies, are tests commonly used before testing them experimentally. Wire material and the cross-section show a great influence in the choice of the geometric variables because the modulus of elasticity (E), modulus of resiliency (R) and the moment of inertia (I). Experimental tests should be preceded by numerical methods because the latter idealize a particular setting, and allows modifying variables.

INTRODUCTION

Sectional springs can be used to retract canines[1,2] as well as to act as part of a segmented arch[3-7] (attraction-springs) that retracts the anterior segment, protract the posterior segment, or both at the same time, depending on the treatment planning (loop position, alpha and beta moments will determine the type of tooth movement)[6-12]. The effectiveness of a certain spring is related to its geometry and the nature of the wire material[13]. Springs must work within an elastic limit, should not harm the adjacent tissues and should perform a force system capable of producing controlled movement of one or more teeth. Alloys which are less resistant to deflection might be displaced more widely, without plastic deformation, and this is a characteristic of non-linear systems[9,14,15]. Among the mechanical properties that characterize spring behavior, spring rate plays an outstanding role, for it allows the clinician to know its load deflection rate[1-5, 7-10]. At first, only gold alloys were used, later were introduced the stainless steel (SS) and cobalt-chromium (Co-Cr) wires. Only the force levels produced after a certain elastic deformity (activation) were studied, considering the force stored per activation unit (spring-gradient), through rudimentary methods, and only dead weights were used. Literature shows that as time went by, the springs started presenting a more complex force system with the advent of titanium-molybdenum alloys. The experimental methods became more sophisticated with the strain gauges[11-16] as well as mathematical models, software programs and elaborated analytical methods (e.g., the finite element method, FEM). FEM was developed around the 1960´s and is widely used in aerospace and engineering industries[3,4,16,17].

The aim of this study was to perform a review of the literature concerning closing loops regarding the methods used for measurements, the mechanical properties (wire material), spring geometry and the resultant force system.

LITERATURE REVIEW

Geometry and methods used for measurements (design study):

Several orthodontic retraction spring geometries[1-50] meant to close spaces in a controlled and systematic way have been studied and developed[1-19]. Experimental and analytical are the most prevalent in the spring´s studies. In the past, Yang and Baldwin[9] compared variations of 0.017 x 0.022-in vertical-loop and Bull-loop made of stainless steel (SS) experimentally through an electro-mechanical device for measuring the spring rate, applied forces and resulting deflections. The springs were also analytically studied through the FEM. Also, Vanderby et al.[18] studied T-loops, L-loops and rectangular-loops (0.010 x 0.021-in) by means of an apparatus capable of measuring vertical force and moments with a linear variable differential transformer (LVDT) and angular transducers, respectively.

According to the authors, the horizontal forces were not measured but obtained by static equilibrium equations. Each loop was centered, and their extremities rigidly fixed at an interbracket distance (IBD) of 7.0 mm. The loops were activated gingivally and occlusally up to 3.0mm. The loops had 6.0 mm height, 7.0 mm length and the active portion was variable. Koenig et al.[11] evaluated T-loop and rectangular-loops experimentally and analytically with a sophisticated mechanical-electronic device used to understand a one-level force system involving forces and moments delivered by the springs. Later in 1982, Burstone[7] studied the T-loop "attraction springs" (0.017 x 0.025-in, titanium-molybdenum, TMA) experimentally using a specially designed force transducer, and then tested clinically the performance of these springs for anterior retraction, posterior protraction and for both anterior and posterior segments. Siatkowski[4,5] developed a new spring design, the Opus-loop (TMA 0.017 x .025-in, and SS wire 0.016 x 0.022-in and 0.018 x 0.025-in); at first, an analytical approach was taken, which was followed by an experimental study using the load cell to measure force and a transducer to measure moments. The characteristics of a modified NiTi and TMA closing loops were verified by Kum et al.[20] considering three different geometries (U-Loop; symmetrical T-loop and asymmetrical T-loop). It was utilized a testing apparatus comprising force and moment transducers to measure the forces and moments resulting from activation and deactivation of closing loops. The authors concluded that NiTi loops produced a lower stress level than TMA loops. Ferreira et al.[8] measured the force system and the interaction between the geometrical parameters developed by a new orthodontic retraction spring (Delta Spring) and found a complex system resulted due to large displacements (activations up to 5mm). A transducer was used for measuring forces and moments, an aluminum structure made up of a cross-shaped beam tied to its internal part, with twelve strain gauges was utilized. The spring design presented delta geometry with a superior helix and gables in their extremities with 0.017 x 0.025-in. Thiesen et al.[24] verified the effect of incorporating gables and helices in T-Loop geometry with 0.017 x 0.025-in and 0.019 x 0.025-in springs, with a transducer device capable of measuring force and moments. They found that T-loops with gable yielded higher M:F ratios and that the incorporation of helicoids seemed to be unnecessary. Coimbra et al.[25] tested seventy-five SS tear-drop loops with 0.019 x 0.025-in, with different heights (6.0mm, 7.0mm and 8.0mm). The loops were designed first in AutoCad software and FEM and then loops were attached to a universal testing machine and tensile tests were performed. Activations were made by 0.5 mm increments until 2.0 mm.

The authors have focused on the forces and moments and they found that the torque angle created after activation (ranged from 0.2° to 1.4°) could explain absence of dimensional symmetry in the prototypes. Loops with 8.0 mm height produced the lower force values during activation. After 2.0 mm of activation a mean value of force was 5.40 N. The highest stress concentrations were located at the loop curvature. Falkner et al.[26] evaluated the effects on the M:F ratio produced by T-loops with various parameters (pre-activation angulation, spring position, height, and the addition of helices) using the FEM and experimental tests and found that varying the spring height, the M:F ratios produced became larger and if the loop's height are increased the intrusive/extrusive forces are modified, also that off-center positioning had a significant impact on the moments, but the incorporation of helices have no clinical significance. Maia et al.[29] used the photoelastic model to evaluate the force system from centralized retraction T-loop springs with helicoids made of stainless steel, also TMA without helicoids. In this study the authors demonstrated that despite the fact that SS T-loops produced higher force magnitudes than that TMA T-loops did, both had the same mechanical characteristics, i.e., the M:F ratio did not change according to the type of alloy used (SS and

TMA) for T-loops at 5 mm of activation and was similar in both sides.

Spring-rate (Spring-gradient; spring-constant; load deflection rate)

Some articles[1,3,4,7-13] use the term spring rate or spring gradient to refer the force (load) necessary for each millimeter of activation or, conversely, to the force released at each millimeter during the deactivation (unloading). One spring whose spring rate is high causes the force magnitude to be altered suddenly, while a spring with low spring rate releases a constant force throughout the entire range of activation. The spring rate is dependent on wire material stiffness (modulus of elasticity, moment of inertia, cross-section and spring design)[13]. The spring-rate is influenced by some factors, such as the wire material of the alloy, the cross-section and the spring design. If the springs present the same design and the same cross-section, the rate will be altered due to the wire material or due to the modulus of elasticity (E). In this way, titanium-molybdenum springs would present advantages over stainless steel and cobalt-chromium alloys, since they present a smaller E[7]. The spring rate must be analyzed according to the spring design once this mechanical property represents the stored force at each activation millimeter (stiffness). Springs whose load-deflection rate is low deliver more constant forces over teeth, so that less damage will occur over periodontal structures. Scientific literature shows spring rates which vary from 33 gf/mm7 to 114 gf/mm in L-loops[9]. In 1982, Burstone[7] studied the effect of composite T-loop (0.018 x 0.017 x 0.025-in) and found a spring rate of 33gm/mm. Gjessing[1] developed a stainless steel 0.016 x 0.022-in spring with a designed ovoid double helix loop showing 160 gm when the double helices were 1.0 mm apart. The spring rate was 45gm/mm. Bench et al.[33] after verifying cobalt-chromium 0.016 x 0.016-in (Blue Elgiloy) springs in a double vertical helical closing loop design, 60mm long, noticed that 2.0 to 3.0 mm activation are necessary to get a load of 100 to 150 gm. The spring rate was 75 gm/mm. Ferreira et al.[15] found 34 gm/mm for delta geometries.

Moment-to-force ratio (M:F)

After this review it was concluded that among the loop's geometries studied no one was capable to produce optimum moment-to-force M:F ratio or capable to produce translation for en-masse tooth movement for a long time. To maintain the desired M:F ratio clinician should at appointment re-insert or fix the gable bends and verify the spring´s dimensions in relation to the interbracket distance as the space closes. Geometrial factors were also explored in many papers reviewed claiming on best results for M:F ratios. Bourauel et al.[21] tested mechanical and geometrical parameters in a Burstone's[7] modified T-loops made of superelastic NiTi alloy (0.016 x 0.022-in) with steel arms (0.017 x 0.025-in). The authors emphasize that even though the T-loops resulted in nearly constant forces, there were differences in the distalizing force and M:F because the springs were strongly influenced by the alloy composition and thermal treatment, in this way the spring should be calibrated individually. Chen et al.[22] verified the effects of dimensional changes on SS T-loops (0.016 x 0.022-in) considering the force system resulting of activations in relation to Mz:Fx ratio. Two groups were studied (measuring Fx, Fy and Mz) one of them without gable bends and heat treatment and another with gable bends (30°) and heat treatment. The authors concluded that even though the heat treatment and gables help to increase the Mz:Fx, they were not enough to produce translation. It is possible to obtain better control of the center of rotation (Crot), in order to avoid stress areas and pressure upon the apical and cervical regions when gable bends are used[27]. Consequently, one can predict the type of movement that is expected, once the M:F relation used is previously known, according to the anchorage requirements expected.

According to Burstone et al.[7], the M:F for central incisors varies depending on the type of movement: (1) Translation movement M:F= 10; (2) Movement around the radicular apex M:F=5; (3) Movement around the crown M:F=12. Raboud et al.[17] found that, for upper canines, when the M:F is 8.5, there will be translation, while there will be inclination around the radicular apex (Crot) when the M:F is below 8.5, and there will be inclination around the dental crown (Crot) when the M:F is above 8.5. Gjessing[1] advocates that the M:F ratio should be about[8-11] to produce bodily movement during canine and anterior teeth. Falkner et al.[26] found that varying the spring height in T loops results in larger M:F as the height increases, while the incorporation of helices at the bends little contributes to alter the mechanical properties. Braun and Marcotte[50] states that, for protraction of the posterior segment (type C extraction site closure) M:F relation should be equal to 10.0 mm at the posterior segment and approximately 13.0 mm at the anterior segment; that would produce translation of the posterior segment and root movement applied to the anterior teeth. In type B, attraction mechanics with a force system applied equally to both segments is expected; thus, the M:F should be approximately 10. Finally, in type A the M:F produces translation on the anterior segment (M:F=10) and root movement on the posterior segment (M:F=13). Those authors believe that translation movements would produce faster movements than root movements; in this way, the necessary anchorage objectives will be reached. Sander[3] points out that the hybrid spring is capable of producing a constant M:F relation of 10.0 mm along 4.0 mm, while the Mz:F (anti-rotation) relation is 3 mm. Besides the M:F ratios some papers[34,35] verified the effects of first- and second-order bends on the force and moments produced by T-loop archwires. It was concluded that gable bends alter the load system; however, the bends produced unpredict table results on the force systems. Also, an analytical approach comparing 2D and 3D analysis was carried out that concluded that the traditional 2D system became inaccurate, if adapted to the 3D system without a modified coordinate system[36].

Force Levels

The literature shows that the usual units used to express force is gf (gram-force), N (Newtons) or cN (centi-Newton). In the International System of Units (SI), the unit of force is N and 1 gf is equal to 0.00980665 N. cN is a decimal fraction of Newton (1 cN = 0.01 N) Regarding the orthodontic literature, the force magnitude can vary depending on the number of teeth considered during movement. Burstone[7] advises that TMA (0.017 x 0.025–in) attraction-springs must be kept below 300 gf to minimize anterior retraction and encourages posterior protraction. Braun et al.[10] describe forces developed by T-loops (TMA, 0.017 x 0.025-in) which vary from 50 gf to 300 gf, where activations at 8.0 mm would generate 300 gf for an en-masse (group of teeth) closure, while at 4.0 mm they would produce 150 gf for the retraction of canines. Sander[3] states that the hybrid spring is capable of generating about 120 gf for retraction and about 20 gf for the extrusive force component. Gjessing[1] advocates a force level varying from 100 to 120 gf to distalize canines without surpass the stress levels of periodontal ligament.

Wire Material (Mechanical properties)

There are some mechanical properties that characterize the spring behavior, as springback (term used to refer how a spring can be activated and return to its neutral position without plastic deformation), spring-rate (stiffness) that is directly proportional to the elastic modulus (E) and stored energy (SE). YS (yield stress), maximum elastic load (MEL), and the moment of inertia (MI) of the cross-section of the wire are mechanical properties that can also characterize the spring´s behavior. Spring-rate plays a major role, for it allows the clinician to know its load deflection rate[7,9] or how much energy can be stored (proportional to modulus of resiliency) was possible after activation. MEL is the highest force that can be applied to a metallic alloy without producing permanent deformation. It is the property that limits the manipulation of the factors responsible for the resistance to deflection, because, if the forces exceed the maximum elastic load, the mechanical stress in the wire exceeds the YS, and it undergoes permanent deformations (also called plastic deformations), modifying its dimensions and elastic properties, consequently, the performance of the loop. If the loop has a very low maximum elastic force, it will easily undergo plastic deformation due to the forces of chewing, or even by the forces generated during its activation. This property responds proportionally to the third power of the diameter of the circular section of a wire, and to the width multiplied by the square of the height of the cross-section in wires of rectangular sections.

The increase in wire length for performing circumvolutions, in order to lower the resistance to deflection, does not change the maximum elastic load. The spring-rate is dependent on the modulus of elasticity (E) (also denoted as Young's modulus) and the spring geometry. E is an index of material's stiffness or rigidity, therefore, the larger this module, the greater the resistance to deflection, that is, the greater the accumulated load per millimeter of activation of the loop of a spring. MI describes the capacity of a cross-section to resist bending, hence, round, square or rectangular cross-sections for the wire provide different values for this property and influence in the spring's behavior. Thus, by choosing materials such as titanium-molybdenum (ß-titanium), whose E is less than that of SS, one can decrease the spring-rate and the resistance to deflection. Titanium-molybdenum alloys have good flexibility and conformability, which makes them elective for retraction springs. These alloys have a YS of approximately 1,280 MPa and an E of 69 GPa, which corresponds to 33% of the E of SS and about 35% of those of chromium-cobalt alloys. The resulting stress from retraction spring activation lead to considerations about its elastic range, that is, how far a spring may be activated without surpassing the YS considering that once it is attained, it will no longer respond satisfactorily The resulting stress from retraction spring activation lead to considerations about its elastic range, that is, how far a spring may be activated without surpassing the YS considering that once it is attained, it will no longer respond satisfactorily. At first, the alloys used were made of gold; later on, stainless steel alloys started being used. In the 1970´s it was introduced the cobalt-chromium (Co-Cr) for use in orthodontics (Blue Elgiloy)[13]. Nowadays, the most commonly used alloys are made up of titanium-molybdenum (ß-titanium). The E of SS wires are similar to Co-Cr. In relation to E for the titanium-molybdenum alloys an 8.7 x 10^6 psi and 9.9 x 10^6 psi are found for those with cross-section of 0.017 x 0.025-in and 0.019 x 0.025-in, respectively. The Co-Cr alloys present around 26 x 10^6 psi, while the SS alloys range from 23 x 10^6 to 26 x 10^6 psi. One advantage of using titanium-molybdenum alloys for retraction springs is the relatively lower forces generated by these wires during activation, e.g., vertical forces (extrusion forces) lead to lower reaction forces than those found in SS and Co-Cr alloys (for same spring geometry), besides providing more flexibility[13].

METHODS

Literature search strategy

It was performed a literature search in MEDLINE through PubMed Eletronic database (www.ncbi.nlm.nih.gov/pubmed) digital archive of biomedical and life sciences journal literature at the U.S. National Institutes of Health, NHI, from 1974 to 2014, using search (subject heading) for “orthodontic AND retraction AND springs” (118 items) and “orthodontic AND closing AND loops” (29 items) summing 147 papers.

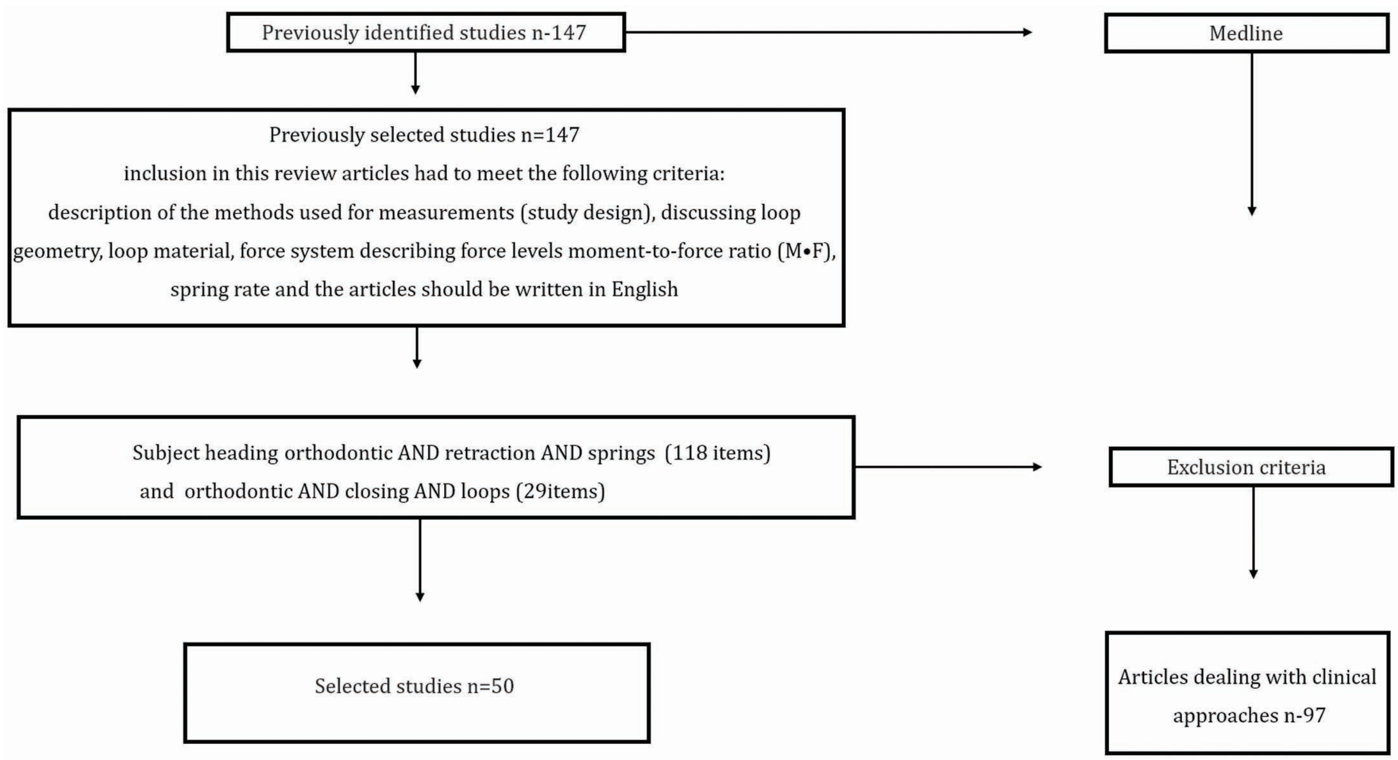

Figure.

For inclusion in this review articles had to meet the following criteria: description of the methods used for measurements (study design), discussing loop geometry, loop material, force system describing force levels, moment-to-force ratio (M:F), spring rate and the articles should be written in English. Articles considering clinical approaches only were discarded. The articles chosen, authors and journals are summarized.

Table - Literature Review. Authors, Wire Material and Spring Geometry

Authors	Wire nature and cross section	Spring Geometry	References
Gjessing (1985)	Stainless Steel (0.016 x 0.022-in)	Double Ovoid	1
Burstone and Koenig (1976)	Stainlees Steel (0.016-in; 0.010 x 0.020-in)	Vertical loop T-loop (with and without helix)	2
Sander (2000)	0.016 x0.022-in	T-loop (hybrid) with coil	3
Siatkowski (1997)	Titanium-molybdenum 0.017 x 0.025-in.	T-loop	4
Siatkowski (1997)	Titanium-molybdenum 0.017 x 0.025-in.	T-loop	5
Rinaldi (1995)	Elastic, coil spring, stainless steel bars with hooks	Segmental spring mechanism prototype	6
Burstone (1982)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	7
Ferreira et al. (2005)	Titanium-molybdenum 0.017 x 0.025-in 0.016 x 0.022-in	Delta Spring	8
Yang and Baldwin (1974)	Stainless-Steel 0.017 x 0.022-in	Bull-loop Vertical-loop	9
Braun et al. (1997)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	10
Koenig et al. (1980)	Stainlees Steel 0.016 x 0.022-in	T-loop and rectangular-loops	11
Solonche et al. (1977)	Stainless steel 0.010x 0.021-in	T-loop L-loop	12
Ferreira (1999)	0.016 x 0.016-in 0.016 x 0.022-in 0.017x 0.025-in 0.018 x 0.025-in 0.019 x 0.025-in Titanium-molybdenum Stainless steel Cobalt-chromium	Double Delta Spring	13
Cavina and Waters (1988)	Multilooped span Average clinical size	Retangular multiloop	14
Ferreira et al. (2013)	Titanium-molybdenum 0.016 x 0.022-in	Delta spring	15
Safavi et al. (2006)	Stainless steel 0.016 x0.022-in	Opus loop, L-loop, T-loop and Vertical helical loop	16
Raboud et al. (1997)	Titanium-molybdenum Stainless steel 0.017 x 0.025-in	T-loop Vertical loop	17
Vanderby et al. (1977)	Stainless steel 0.010x 0.021-in	T-loop L-loop	18
Thiesen et al. (2013)	Titanium-molybdenum Stainless steel 0.017 x0.025-in 0.019 x 0.025-in	Tear drop loops with and without helix	19

Authors	Wire nature and cross section	Spring Geometry	References
Kum et al. (2004)	Titanium-molybdenum NiTi	U-loops, T-loops, asymmetrical T-loops (TMA) T-loop (NiTi) X-loop (TMA)	20
Bouaruel et al. (1997)	Super-elastic NiTi (0.016 x 0.022-in) +SS arms (0.017 x0.025-in)	T-loop	21
Chen et al. (2000)	Stainless steel 0.016 x 0.022-in	T-loop	22
Darendeliler et al. (1997)	Stainless steel 0.006 x 0.020-in	Drum spring	23
Thiesen et al. (2005)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	24
Coimbra et al. (2007)	Stainless steel 0.019 x 0.025-in	Tear drop loop	25
Faulkner et al. (1989)	Titanium-molybdenum 0.017 x 0.025-in	T-loop with and without helices	26
Martins et al. (2008)	Titanium-molybdenum 0.017 x 0.025-in	T-loop peractivated by curvature and preactivated concentrated bends	27
Caldas et al. (2011)	Titanium-molybdenum 0.017 x 0.025-in	T-loop peractivated by curvature and preactivated with V- bends	28
Maia et al. (2011)	Titanium-molybdenum 0.017 x 0.025-in	T-loop with and without helicoids	29
Rose et al. (2009)	Titanium-molybdenum Nickel-titanium	T-loop	30
Techalertpaisarn et al. (2013)	Stainless steel 0.016 x 0.022-in	Opus loop, L-loop (with and withou coil) and T-loop	31
Techalertpaisarn et al. (2013)	Stainless steel 0.016 x 0.022-in	T-loop, vertical-loop and L-loop	32
Xia et al. (2013)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	33
Katona et al. (2006)	Stainless steel 0.016 x 0.022-in	Triangular loop	34
Katona et al. (2014)	Stainless steel 0.016 x 0.022-in	T-loop	35
Katona et al. (2014)	Stainless steel 0.016 x 0.022-in	T-loop	36
Chen et al. (2007)	Stainless steel 0.016 x 0.022-in	Triangular loop	37
Kojima and Fukui (2012)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	38
Hoenigl et al. (1995)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	39
Caldas et al. (2011)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	40

Authors	Wire nature and cross section	Spring Geometry	References
Menghi et al. (1999)	Titanium-molybdenum 0.017 x 0.025-in	L loop, rectangular-loop and T-loop	41
Eden et al. (1994)	Stainless steel 0.016 x 0.022-in	PG spring	42
Chen et al. (2010)	Stainless steel 0.016 x 0.022-in	T-loop archwires	43
Lim et al. (2008)	Nickel-Titanium, Titanium-molybdenum 0.017 x 0.025-in	T-loop	44
Geramy et al. (2012)	Stainless steel 0.15 x 0.21-in	Vertical loops with modifications in design	45
Odegaard et al. (1996)	NiTi and stainless steel wires, 0.018 x 0.025-in; 0.016 x 0.022-in (Tear drop loop)	T-loop, Reverse helical closing loop, Bull-loop, Key-hole loop and Tear drop loop	46
Blaya et al. (2009)	Stainless steel Titanium-molybdenum 0.017 x 0.025-in 0.016 x 0.016-in	Tear drop loop and circle-shaped loop	47
Kumar et al. (2009)	Stainless steel (0.016 x 0.022-in) Titanium-molybdenum (0.016 x 0.022-in)	PG spring (SS); T-loop (TMA)	48
Manhartsberger (1989)	Titanium-molybdenum 0.016 x 0.022-in 0.017 x 0.025-in	T-loop	49
Braun and Marcotte (1995)	Titanium-molybdenum 0.017 x 0.025-in	T-loop	50

RESULTS

Literature search results

From the 147 papers resulted from Pubmed, 50 were chosen because they have met the selection criteria, i.e., they directly deal with the subject "retraction springs" or "orthodontic closing loops". Figure Overall, many works seek not only improve the knowledge about the force system (Fx; Fy; Mz; Mx; My and M/F) generated by the springs and the forces levels, but also better understand the mechanical behavior by means of experimental tests, photoelastic studies, holographic studies, software programs or mathematical studies.

DISCUSSION

In this review it was found that the design of a closing loop involves the study of the geometric parameters such as the helicoids, the spring height, the width, the length wire, the incorporation of gables at their ends (which generate moments) and the IBDs[7,12,13,24-31,34-36]. Also, the material nature and the cross-section used have a great influence in the design and in the choice of the geometric variables because the E[13] and its MI. Experimental tests should be preceded by analytical studies because the latter allow idealize a particular setting, modifying variables, if necessary, in addition to being cheaper than experimentation. Nevertheless, according to some authors[11], experimental methods have the advantage of being able to take into account variables that cannot be included in analytical models. Several spring geometries are now available and have been tested experimentally[3,4,7-9,11,13,15,18,22,37-42,45,48-50], software programs[27,28] and tested also numerically using the FEM[6,9,11,25,27,30,36,43,44,46]. Experimental methods have been employed to reproduce analytical models, allowing for a better understanding of their behavior. It was also concluded that calibrating the models and refining the experiments can provide useful insights into the spring mechanical behavior before they are clinically applied[3,4,7,15].

An important study[29] showed that the state of stress, compression and tension can be also measured by photoelastic methods even in irregular or asymmetrical geometries. A sensitive photoelastic resin is used and its critical stress points are visualized after an imposed deformation. The principle is that of birefringent materials (a property of some solids called isotropic, that after stress become doubly refracting) which experiment two refractive indices after a ray of light passes through a transparent plastic and the resulting polarized beams. An optical device with polarized light is necessary to make the measurements of resulting beams that are viewing as colored fringes of varying sensitivity. Holographic studies are also used in Orthodontics. In the holographic studies (laser holography a noninvasive method can be used for determine displacements) the three-dimensional image is encoded in a two-dimensional surface where the object is reconstructed in light. Kumar et al.[48] studied holographically the behavior of T-loops, closed and open coil springs and the PG spring. The aim was to know the estimate magnitude and direction of initial displacement of the canine. The authors found that PG springs produced the highest initial displacement followed by open coils springs, closed coil springs and finally T-loop springs, but maximum tipping was produced by open coil springs, PG springs, closed coli springs and T-loop springs, respectively. PG springs is preferred for higher magnitude of displacement and closed coil springs for moderate displacement and tipping. T-loop is preferred for minimal tipping.

The FEM is a numerical method for solving complex problems (engineering and mathematical) or complex structures (structural analysis) in parts called fini. The FEM have been used more frequently and with new versions is a computer-aided 3-D models, an important tool to verify the internal stress behavior of a material. The basic idea of the method is to split the body or domain studied into sub-regions, the FEs. The equations pertaining to each element are then joined, so that continuity is preserved, and one global equation is obtained to represent the entire body based on the mathematical equations that govern the phenomenon studied for each sub-region (FE). In the static analysis of stress and strains, the equation that represents the body is given by $[K]\{u\}=\{F\}$, where $[K]$ is the stiffness matrix, $\{u\}$ is the nodal displacement vector, and $\{F\}$ is the nodal force vector. After finding the nodal displacements $\{u\}$ through the solution of the algebraic system shown in equation, the stresses and efforts on the body may be evaluated[15].

The springs should have a geometry such that when it is activated it is sustained itself between their horizontal edges (legs) which usually cross themselves in the activation, thus it will be more difficult to incline to the side of the oral mucosa or gingival[8,15]. Therefore, the height of the spring should be kept between 6 to 10 mm maximum[22]. Also, symmetric springs produce very slight vertical forces compared to non-symmetric ones. It was also, pointed out that increasing the vertical and horizontal dimensions it tends to decrease the moment magnitude in stainless steels T-loops and incorporating gables increase the load deflection rate and the same happens to the M:F ratio[20]. The moments will have the same magnitudes since they present the same gable bends in both extremities. Literature shows that there is no ideal spring and there are many different geometries available, hence, it lies to the clinician to choose the one he/she better master and the one that can produce appropriate movement within a M:F ratio almost constant, with easy activation and that produces light forces, but which are able to move the teeth efficiently and do not cause ulcerations in the patient. Nowadays, titanium-molybdenum (ß-titanium) springs are preferred because they present low force-deflection ratios compared to SS and Co-Cr, and because they have a good spring-rate. It is preferable they are built using rectangular cross-sections because better vertical, horizontal and transversal controls are possible, to avoid bucco-gingival inclination that can occurs during the activation.

CONCLUSIONS

- One of the difficulties concerning the use of springs is related to the fact that they need a previous study of the force system developed by them that allows the springs to be preactivated more accurately.
- Springs made of ß-titanium alloys produce lower spring-rates.
- These springs can be part of a segmented or of a sectional approach, depending on the treatment planning.
- Springs should be correctly fashioned to avoid harm tissues.
- Some methods have been implemented in order to better understand the performance of the springs. FEM, holographic and photoelastic studies, are techniques commonly used before testing them experimentally.

- Material nature and the cross-section show a great influence in the design and in the choice of the geometric variables because the modulus of elasticity (E) and the moment of inertia (MI).
- Experimental tests should be preceded by analytical-numerical studies because the latter allow idealize a particular setting, and allows modifying variables, if necessary.

There is no conflict of interest.

OJN

REFERENCES

1. Gjessing P. Biomechanical design and clinical evaluation of a new canine retraction spring. Am J Orthod 1985;87(5): 353-62.
2. Burstone C J, Koenig H A. Optimizing anterior and canine retraction. Am J Orthod. 1976; 70(1):1–20.
3. Sander FG. Biomechanical Investigation of the hybrid retraction spring. J Orofac Orthop. 2000; 61(5): 341-51.
4. Siatkowski RE. Continuous archwire closing loop design, optimization, and verification. Part I. Am J Orthod Dentofacial Orthop. 1997; 112(4): 393-402.
5. Siatkowski RE. Continuous arch wire closing loop design, optimization, and verification. Part II, Am J Orthop Dentofacial Orthop; 1997.112(5): 487-495.
6. Rinaldi TC. Johnson BE. An analytical evaluation of a new spring design for segmented space closure. Angle Orthod. 1995; 65(3): 187-198.
7. Burstone CJ. The segmented approach to space closure. Am J Orthod. 1982;82(5): 361-378.
8. Ferreira MA, Oliveira FT, Ignácio AS, Borges PC. Experimental force definition system of a new orthodontic retraction spring. Angle Orthod. 2005; 75(3): 334-343.
9. Yang TY, Baldwin JJ. Analysis of space closing springs in orthodontics. J Biomech. 1974; 7(1): 21-28.
10. Braun S, Sjursen RC, Legan HL. On the management of extraction sites. Am J Orthod Dentofacial Orthop. 1997;112(6): 645-655.
11. Koenig HA, Vanderby R, Solonche DJ, Burstone CJ. Force system from orthodontic appliance: An analytical and experimental comparison. Trans ASME. 1980; 102(4): 294-298.
12. Solonche DJ, Burstone, CJ, Vanderby, RA device for determining the mechanical behavior of orthodontic appliances. IEEE Trans Biom Eng. 1977; 24(6): 538-539.
13. Ferreira MA. The wire material and cross-section effect on double delta closing loops regarding load and spring rate magnitude: An in vitro study. Am J Orthod Dentofacial Orthop. 1999; 115(3): 275-82.
14. Cavina RA, Waters NE. The behaviour of multiple loops in torsion. Br J Orthod. 1988; 15(3):149-56.
15. Ferreira MA, Assumpção R, Luersen MA, Borges PC. Mechanical behaviour of a prototype orthodontic retraction spring: a numerical-experimental study. Eur J Orthod. 2013; 35(4):414-20.
16. Safavi MR, Geramy A, Khezri AK. M/F ratios of four different closing loops: 3D analysis using the finite element method (FEM). Aust Orthod J. 2006; 22(2):121-6.
17. Raboud DW, Faulkner MG, Lipsett AW, Haberstock, DL. Three-dimensional effects in retraction appliance design. Am J Orthop Dentofacial Orthop; 1997. 112(4): 378-392.
18. Vanderby R, Burstone CJ; Solonche DJ; Ratches JA. Experimentally Determined Force Systems from Vertically Activated Orthodontic Loops. The Angle Orthod. 1977; 47(4): 272–279.
19. Thiesen G, Shimizu RH, do Valle CV, do Valle-Corotti KM, Pereira JR, Conti PC. Determination of the force systems produced by different configurations of tear drop orthodontic loops. Dental Press J Orthod. 2013; 18(2):19. e1-18.
20. Kum M, Quick A, Hood JA, Herbison P. Moment to force ratio characteristics of three japanese NiTi and TMA closing loops. Aust Orthod J. 2004; 20(2):107-114.
21. Bourauel C, Drescher D, Ebling J, Broome D, Kanarachos A. Superelastic nickel titanium alloy springs—an investigation of force systems. Eur J Orthod. 1997; 19(5):491–500.
22. Chen J, Markham DL, Katona TR. Effects of T-loop geometry on its force and moments. Angle Orthod; 2000; 70(1): 48-51.
23. Darendeliler MA, Darendeliler H, Uner O.The drum spring (DS) retractor: constant and continuous force for canine retraction. Eur J Orthod. 1997; 19(2):115-30.
24. Thiesen G, Rego MNN, Menezes LM, Shimizu RH. Force system yielded by different designs of T-loop. Aust Orthod J. 2005; 21(2):103-110.
25. Coimbra MER, Penedo ND, Gouvêa, JP, Elias CN, Araújo MTS, Coelho PG. Mechanical testing and finite element analysis of orthodontic tear-drop loop. Am J Orthop Dentofacial Orthop. 2007.133:188.e9-188.e13).

26. Faulkner MG, Fuchshuber P, Haberstock D, Mioduchowski A. A parametric study of the force/moment system produced by T-loop retraction springs. J of Biomech. 1989; 22(6-7) :637-647.
27. Martins RP, Buschang PH, Viecilli R, Santos-Pinto A. Curvature versus V-bend in a group B titanium T loop spring. Angle Orthod. 2008; 78(3):517-523.
28. Caldas SGFR; Martins RP; Galvão MR; Vieira CIV; 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(2): e53-e58.
29. Maia LGM, Maia MLM, Monini AC, Vianna AP, Gandini Jr LG. Photoelastic analysis of forces generated by T-loop springs made with stainless steel or titanium-molybdenum alloy. Am J Orthod Dentofacial Orthop. 2011; 140(3): e123-e128.
30. 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 Dento Facial Orthop. 2009; 135(6):757-63.
31. Techalertpaisarn P, Versluis A. How do mechanical responses at closing loop ends vary when loop position changes? A systematic analysis of vertical, T-loops, and L-loops. Oral Sci Intern. 2013; 10(2):58-64.
32. Techalertpaisarn P, Versluis A. Mechanical properties of Opus closing loops, L- loops, and T-loops investigated with finite element analysis. Am J Orthod Dento Facial Orthop. 2013; 143(5):675-683.
33. Xia Z, Chen J, Jiangc F, Viecilli RF, Liu SY. Load system of segmental T-loops for canine retraction. Am J Orthod Dento Facial Orthop. 2013; 144(4):548-56.
34. Katona TR, Le YP, Chen J. The effects of first- and second-order gable bends on forces and moments generated by triangular loops. Am J Orthod Dentofacial Orthop. 2006; 129(1):54-59.
35. Katona RK, Isikbay SC, Chen J. Effects of first-and second-order gable bends on the orthodontic load systems produced by T-loop archwires. Angle Orthod. 2014; 84(2):350-357.
36. Katona RK, Isikbay SC, Chen J. An analytical approach to 3D orthodontic load systems. Angle Orthod. 2014; 84(5): 830-838.
37. Chen J, Bulucea I, Katona TR, Ofner S. Complete orthodontic load system on teeth in a continuous full archwire:The role of triangular loop position. Am J Orthod Dentofacial Orthop. 2007; 132(2):143. e1-143.e8.
38. Kojima Y, Fukui H. Numerical simulations of canine retraction with T-loop springs based on the updated moment-to-force ratio. Eur J Orthod. 2010; 34(1):10-18.
39. Hoenigl KD, Freudenthaler J, Marcotte MR, Blanteon H-P. The centerd T-loop- a new way of preactivation. Am J Orthod Dentofacial Orthop. 1995; 108(2):149-53.
40. Caldas SGFR, Martins RP, Viecilli RF, Galvão MR, Martins LP. Effects of stress relaxation in beta-titanium orthodontic loops. Am J Orthod Dentofacial Orthop. 2011; 140: e85-e92.
41. 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(1): 49-57.
42. Eden JD, Waters NE. An investigation into the characteristics of the PG canine retraction spring. Am J Orthod Dentofacial Orthop. 1994; 105(1):49-60.
41. Bourauel C, Drescher D, Ebling J, Broome D, Kanarachos A. Superelastic nickel titanium alloy retraction springs: an experimental investigation of force systems. Eur J Orthod. 1997; 19(5):491-500.
43. Chen J, Isikbay SC, Brizendine EJ. Quantification of three-dimensional orthodontic force systems of T-loop archwires. Angle Orthod. 2010; 80(4):754-8.
44. Lim Y, Quick A, Swain M, Herbison P. Temperature effects on the force, moments and moment to force of nickel-titanium and TMA symmetrical T-loops. Angle Orthod. 2008; 78(6):1035-42.
45. Geramy A, Retrouvey JM, Shalchi M, Salehi H. Loop position in anterior retraction arch wire and its effects on the produced forces: 3d analysis using finite element method. International Journal of Clinical Dentistry. 2012; 5(2):121-30.
46. Øodegard J, Meling T, Meling E. The effects of loops on the torsional stiffnesses of rectangular wires: An in vitro study. Am J Orthod Dentofac Orthop. 1996; 109(5):496-505.
47. Blaya MB, Westphalen GH, Guimaraes MB, Hirakata LM. Evaluation of tensile strength of different configurations of orthodontic retraction loops for obtaining optimized forces. Stomatologija. 2009;11(66-69):66-9.
48. Kumar YM, Ravindra NS, Balasubramaniam MR. Holographic Analysis of the initial canine displacement produced by four different retraction springs. Angle Orthod. 2009; 79(2):368-372.
49. Manhartsberger C, Morton JY, Burstone CJ. Space closure in adult patients using the segmented arch technique. Angle Orthod. 1989; 59(3):205–210. 13.
50. Braun S, Marcotte MR. Rationale of the segmented approach to orthodontic treatment. Am J Orthod Dentofac Orthop. 1995; 108(1):1-8.