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M/F ratios of four different closing loops: 3D analysis using the finite element method (FEM)

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Background: It has been claimed that the opus loop is capable of delivering a constant and optimum M/F ratio without the need for gable bends.

Objective: To compare the forces, moments and moment/force (M/F) ratios of the opus loop, L–loop, T–loop and vertical helical closing loop (VHC loop) in a segmented arch with the finite element method (FEM).

Methods: The FEM was used to compare 3D models of closing loops in rectangular (0.016 x 0.022 inch) stainless steel wire. The L–, T– and VHC loops were designed with and without preactivation bends. The opus loop had no preactivation bends. The T–Loops were 10 mm in height and 10 mm in length. The horizontal and vertical forces, the moments and the M/F ratios at the alpha and beta ends were recorded at 0, 0.1, 0.4, 0.7 and 1 mm intervals.

Results: The highest horizontal and vertical forces were produced by the L–loop (with and without preactivation bends) and in most cases the lowest forces were produced by the VHC loop. Loops with preactivation bends produced marked changes in the M/F ratio and loops without preactivation bends low, but relatively constant, M/F ratios over the full range of activation. Of the loops modelled without preactivation bends the opus and T-loop had the highest M/F ratios (7.20 – 7.67 mm at the anterior ends).

Conclusion: Stainless steel opus and T–loops without preactivation bends had constant M/F ratios, but both loops failed to deliver the optimum M/F ratio of 10:1.

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Introduction

Closing loops have the ability to move teeth without friction and, theoretically, have the ability to translate a tooth. Two important characteristics of closing loops used for orthodontic space closure are the moment/force (M/F) ratio and the load-deflection (F/D) rate.[1,2] Although these characteristics are influenced by the amount of activation they are useful in defining the mechanical properties of a retraction spring.[3] The vertical helical closing loop (VHC), T- and L-loops have low inherent M/F ratios and large F/D rates. It has been claimed that the opus loop without gable bends has a constant and optimum M/F ratio in a continuous system.[4]

Many stainless steel closing loops have a large F/D rate and the M/F ratio changes rapidly as a loop activates and deactivates. As a result the teeth move through cycles of controlled crown tipping, translation and root movement. Springs made from materials with a lower Young's modulus than stainless steel may pass through fewer cycles for a given amount of space closure. Because traditional stainless steel closing loops are not capable of delivering an ideal M/F ratio of 10:1 clinicians generally add residual moments, in the form of gable bends, to reduce the amount of crown/root tipping. It is difficult to estimate the correct amount of residual moment needed in each clinical situation. Obviously, it would be an advantage to have a loop capable of delivering the ideal M/F ratio over the full range of activation so that cycles of tipping and translation can be avoided.

Specific designs of closing loops can be made, tested and compared in a laboratory setting or modelled by computer software, such as finite element analysis. The finite element method (FEM) is a mathematical

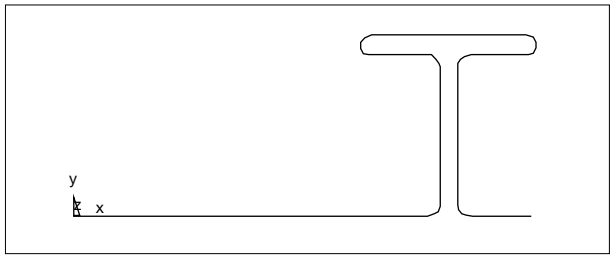


Figure 1a

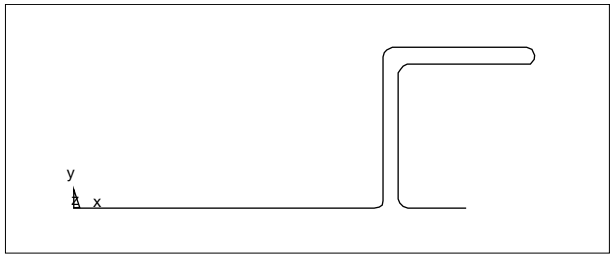


Figure 1b

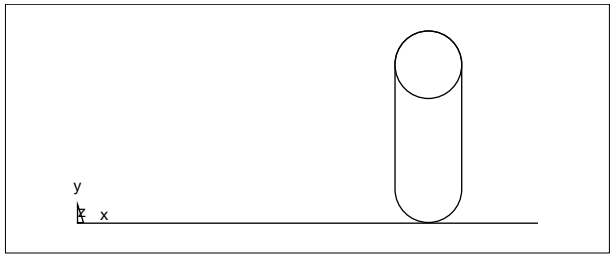


Figure 1c

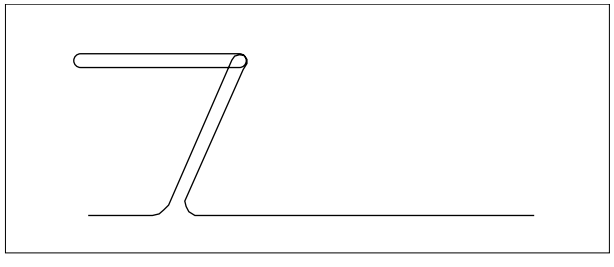


Figure 1d

Figure 1. Loops without preactivation bends.
a) T–loop;
b) L–loop;
c) Vertical helical closing loop;
d) Opus loop.
Loops with a preactivation bend.
e) T–loop;
f) L–loop;
g) Vertical helical closing loop.
Note. The anterior (alpha) ends of the T–, L– and VHC loops are on right side. The anterior (alpha) end of the opus loop is on the left side.

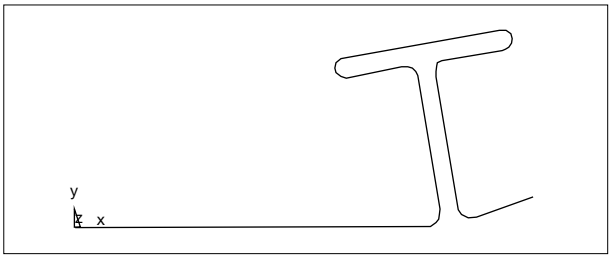


Figure 1e

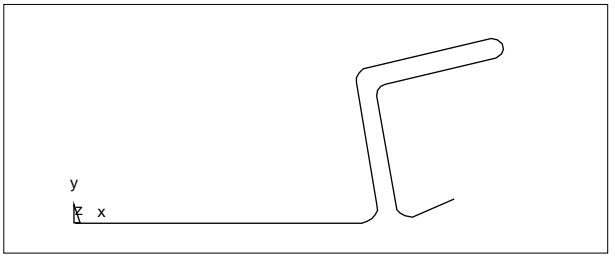


Figure 1f

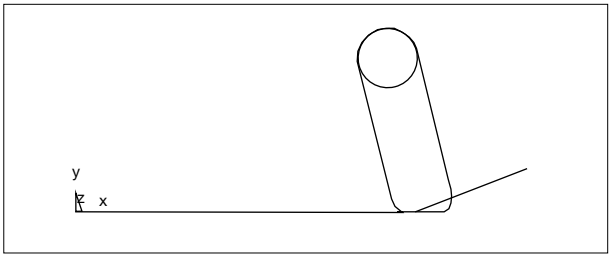


Figure 1g

method in which a complex system of keypoints or nodes are located on an accurate drawing of the structure to be modelled. From each keypoint an element extends to each adjacent keypoint forming a mesh of vectors. The mesh is programmed to 'contain' the material and structural properties that define how the structure will react under certain loading conditions. For example, the behaviour of an orthodontic spring can be modelled by beam element theory. Young's modulus and Poisson's ratio are used to define the form and mechanical properties of the spring material. FEM modelling can also be 2D or 3D and the system can be programmed to behave linearly or non-linearly. The latter can account for plastic deformation in the structure under examination. The FEM has been used to answer a number of different clinical problems in orthodontics and prosthodontics.[5–14]

In this study we aim to use the FEM to model and compare the forces, moments and M/F ratios of three loops (L-loop, T-loop, and VHC loop) with and without preactivation bends and the opus loop without preactivation bends.

Material and methods

The opus, T-, L- and VHC loops for 3D analysis with the FEM were assumed to be made of 0.016 inch × 0.022 inch rectangular stainless steel wire.[1,4–6] A 3D beam element was used to construct the models. The T-, L- and VHC loops were modelled with and without preactivation bends, and the opus loop without preactivation bends (Figure 1). The L-, T-, and opus loops were 10 mm in height and 10 mm in length. The diameter of the VHC-loop was set at 3.5 mm. The model was designed with preactivation bends 10 degrees on both sides of the T-, L-, and VHC loops. In the loops with preactivation bends one 10 degree gable bend was placed between the horizontal posterior segment and the loop and a second 10 degree gable bend was placed between the loop and the anterior segment.

Accurate diagrams of the loops were prepared and the keypoints marked on each diagram. The keypoints were used to transfer the exact geometry of each loop to ANSYS Version 5.71 (ANSYS Inc. Southpointe, Canonsburg, PA, USA) installed in a Pentium IV personal computer.

The cross-sectional area of stainless steel orthodontic wire (0.016 inch x 0.022 inch) was then calculated. The Young's modulus of the stainless steel wire was assumed to be 18×10^6 gr/mm^2 and the Poisson ratio was equal to 0.3. The boundary conditions were defined so that the terminal node in the alpha segment (anterior) was restrained (*i.e.* it was not able to move in the X, Y or Z axes, and it was not able to rotate around these axes). The terminal node of the beta segment (posterior) was restrained in a similar way to the alpha segment, except that it was free to move along the horizontal leg of the posterior segment. This movement simulated the wire sliding distally through a molar tube. The first step was to determine the forces, moments and the M/F ratios of each loop during activation and deactivation (0, 0.1, 0.4, 0.7 and 1 mm) without preactivation bends. The next step was to determine the forces, moments and M/F ratios of the L-loop, T-loop and VHC loop during activation and deactivation with 10 degree preactivation bends.

Loops without preactivation bends were inserted into each model in turn and modelled by defining a fixed point at the terminal node (*i.e.* the alpha position), by displacing the node at the other end (*i.e.* the beta position) and by deriving the forces and moments produced at the terminal nodes. Insertion of a loop with a preactivation bend in mouth contains steps which are performed simultaneously, but have to be defined separately in finite element analysis. The following procedure was used for loops with a preactivation bend:

1. A fixed point at the terminal node (beta position) was defined. This simulated the insertion of the wire into the molar tube.
2. The alpha end was displaced so that the alpha end node remained at the same vertical level as the bracket slot. This simulated displacement of the anterior part of the loop incisally so as to keep it level with the bracket slot. The alpha segment of the loop crossed the bracket slot at an angle, which is directly related to the amount of the preactivation.
3. To simulate engagement of the wire in the brackets a rotation was added to the alpha end until the horizontal leg was completely horizontal and collinear with the bracket slot.
4. The exact displacement at the alpha end node was recorded to obtain information about the amount of cross-over produced when the wire was 'engaged'. This point was the neutral position for all loops (zero force instant) and the starting point for all activations.
5. The different amounts of loop activation were then applied sequentially and recorded.

The analyses were performed on a Pentium IV personal computer by ANSYS Version 5.71. The output data for the forces and the moments were assessed at the terminal nodes.

Results

The horizontal and vertical forces at each loop, the moments at the alpha (anterior) and beta (posterior) ends, and the M/F ratios at both ends when the loops were activated 0, 0.1, 0.4, 0.7 and 1 mm are given in Table I. The forces are given in grams (grf), the moments in gram millimetres (g. mm) and the M/F ratios in millimetres (mm).

Table I. Force and moment data for four different closing loops in 0.016 inch x 0.022 inch stainless steel wire, with and without preactivation bends.

	Loops							
	With preactivation			Without preactivation				
	VHC loop	T-loop	L-loop	VHC* loop	T-loop	L-loop	Opus loop	Activation (mm)
Horizontal force	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
	14.1	14.4	21.2	13.8	12.9	18.7	10.9	0.1
	56.7	57.5	84.7	55.7	51.9	74.5	44.1	0.4
	99.6	100.4	147.7	98.0	90.8	129.7	77.7	0.7
	142.8	143.0	210.0	140.7	129.6	184.0	111.8	1.0
Vertical force	47.6	54.6	-32.5	0.0	0.0	0.0	0.0	0.0
	51.5	59.7	41.3	4.0	4.6	7.5	3.7	0.1
	63.0	74.5	62.7	16.0	18.3	29.9	14.8	0.4
	74.1	89.0	92.7	27.6	31.7	51.7	25.7	0.7
	84.8	103.0	117.7	38.8	44.7	72.8	36.3	1.0
M α	1006.9	944.7	811.3	0.0	0.0	0.0	0.0	0.0
	1086.5	1024.4	942.6	83.0	93.6	130.5	84.1	0.1
	1324.9	1336.3	1345.6	331.6	375.0	526.4	337.8	0.4
	1562.7	1631.6	1761.1	579.2	658.1	927.8	593.1	0.7
	1799.6	1928.1	2187.5	825.6	941.9	1333.4	849.8	1.0
M β	72.8	318.9	-68.0	0.0	0.0	0.0	0.0	0.0
	87.0	343.6	3.7	15.5	23.2	59.9	18.5	0.1
	128.9	416.1	214.6	61.1	91.5	234.8	73.2	0.4
	169.6	487.0	419.0	104.6	157.7	401.9	126.3	0.7
	208.9	555.3	616.5	145.9	221.7	561.2	178.0	1.0
M/F α				0.0	0.0	0.0	0.0	0.0
	76.7	72.2	44.4	5.9	7.2	6.9	7.6	0.1
	23.3	23.2	15.8	5.9	7.2	7.0	7.6	0.4
	15.6	16.2	11.9	5.9	7.2	7.1	7.6	0.7
	12.5	13.4	10.4	5.8	7.2	7.2	7.6	1.0
M/F β				0.0	0.0	0.0	0.0	0.0
	6.1	23.8	6.8	1.1	1.8	3.2	1.7	0.1
	2.2	7.2	2.5	1.1	1.7	3.1	1.7	0.4
	1.7	4.8	2.8	1.1	1.7	3.1	1.7	0.7
	1.4	3.8	2.9	1.1	1.7	3.1	1.6	1.0

* Vertical helical closing loop

Loops without preactivation

The horizontal and vertical forces exerted by the L-loop during activation were higher than the forces calculated for the VHC loop and T-loop. The horizontal forces produced when L- and opus loops were activated 1 mm were 184 and 111.8 grf respectively, and the vertical forces produced by these loops at 1 mm activation were 72.8 and 36.3 grf respectively. The output data showed that the VHC loop had the smallest moments, and anterior (alpha) and posterior

(beta) M/F ratios. The posterior M/F ratios for the L-loop were the highest, while the anterior M/F ratios of the T-, L- and opus loops were more-or-less similar. The anterior M/F ratios for the VHC loop, although consistent over the full range of activation, were less than the M/F ratios obtained for other three loops. The sign (+/-) of the anterior and posterior moments in all loops was the same.

Loops with preactivation

Adding a gable bend produced higher horizontal and vertical forces in all loops. A vertical force was present at zero activation in all loops. The horizontal and vertical forces in the L-loop were again higher than the forces in the VHC and T-loops. Gable bends produced higher moments and, therefore, higher anterior and posterior M/F ratios, especially for small activations. The VHC, T- and L-loops did not maintain a constant M/F ratio over the full range of activation and deactivation.

Discussion

We used the FEM to compare the forces, moments and M/F ratios exerted by four different stainless steel closing loops used in clinical orthodontics. The results indicated that opus loop had the lowest F/D rate, presumably because the geometry of this loop incorporated more wire than the other loops tested. The T- and L-loops used a similar amount of wire, but gave quite different results: the F/D rate of the L-loop was higher that the F/D rate of the T-loop. The opus and T-loops without preactivation bends had rather similar mechanical behaviours. Ten degree preactivation bends changed the vertical and horizontal forces calculated for the L-, T- and VHC loops and, as a result, the M/F ratios.

When working with simple structures it is often possible to obtain a precise answer, but for complex structures a numerical method is generally required. The FEM is a powerful numerical approach able to solve complex elasticity and structural problems in structures, such as orthodontic springs. The method has been used to study a number of clinical and basic problems.[4–14] One of the problems in working with FEM is that the output data is derived from large computer files, which need a scientific background and experience in both medical and mechanical fields. There are several limitations in the method. Firstly, it is not possible to enter data for problems over long periods of time, such as modelling the motion of a tooth subjected to an orthodontic load over a month. Secondly, some permanent (plastic) deformation can occur in loops when they are removed in periodic checks. We did not consider this kind of deformation.

Our data for the loops with preactivation bends agrees with published data.[15–17] The VHC loop showed the lowest M/F ratio, which we attribute to the lack of a horizontal section, and the opus loop had a slightly better anterior M/F ratio (7.6 mm) than the T- and L-loops (7.26 and 7.2 mm respectively). We attribute the differences in the M/F ratios calculated for the opus, T- and L-loops to the different loop geometries.

In agreement with others the M/F ratios of the loops without preactivation bends were more-or-less constant over the full range of activation.[16,17] Adding preactivation bends to the loops resulted in high M/F ratios at 0.1 mm activation, which fell sharply as the activation increased from 0.4 to 1 mm. In accordance with Raboud *et al.*[16] preactivation of the VHC loop doubled the anterior M/F ratio when the loop was activated 1 mm.

The posterior M/F ratios in the loops without preactivation bends were invariably low and independent of the amount of activation. This behaviour in a closing loop will result in anchorage loss. Incorporating preactivation bends increased M/F ratios and, theoretically, should reduce any anchorage loss. In the present work we did not examine the effects of different wire sizes, loop dimensions and inter-bracket distances on the output data. However, others[4,15,17] have reported that high force outputs occur when the wire size is increased (the anterior and posterior M/F ratios remain unchanged) and smaller loops also increase the force and stiffness dramatically, but reduce the M/F ratios. Reducing the inter-bracket distance does not affect the horizontal or vertical forces or the anterior M/F ratio, but the posterior M/F ratio gradually falls to zero.

The opus loop has several advantages over other loops for closing spaces. It has an acceptable M/F ratio and it does not need any preactivation bends, thus making it possible to obtain the desired tooth movement and to provide periods for the periodontium to recover.

Conclusions

Adding preactivation bends to the VHC, T- and L-loops increased the M/F ratios to acceptable levels at 1 mm activation, but there was a dramatic increase in the M/F ratio as the loops deactivated to 0.1 mm. The results indicate that loop activation should be maintained to avoid applying high M/F ratios. Stainless steel opus and T-loops without preactivation bends had constant M/F ratios (7.6 and 7.2 mm respectively) over the full range of activation, but both loops were below the optimum M/F ratio of 10:1.

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References

1. Nanda R. Biomechanics in clinical orthodontics. Philadelphia: W.B. Saunders Co.1997: Chapters 1,10 and 7, p. 188.
2. Lewis PD. Canine retraction. Am J Orthod 1970;57: 543–61.
3. Ziegler P, Ingervall B. Optimizing anterior and canine retraction. Am J Orthod Dentofacial Orthop 1989;95:95–9.
4. Siatkowski RE. Continuous archwire closing loop design, optimization and verification, parts I and II. Am J Orthod Dentofacial Orthop 1997;112:393–402 and 487–95.
5. Geramy A. The same amount of CRes displacement in various orthodontic tooth movements while the applied force remained constant: 3D analysis using finite element method. J Dentistry, Shiraz University of Medical Sciences 2002;3:59–65.
6. Geramy A, Sharafoddin F. Abfraction: 3D analysis by means of the finite element method. Quintessence Int 2003;34: 526–33.
7. Geramy A. Moment to force ratio and the center of rotation: three-dimensional analysis by means of the finite element method. J Dentistry, Shiraz University of Medical Sciences 2002;1:26–34.
8. Geramy A. Alveolar bone resorption and the center of resistance modification (3D analysis by means of the finite element method). Am J Orthod Dentofacial Orthop 2000;117: 399–405.
9. Geramy A. Initial stress produced in the periodontal membrane by orthodontic loads in the presence of varying loss of alveolar bone: a three-dimensional finite element analysis. Eur J Orthod 2002;24:21–33.
10. Geramy A. Stress tensor modification in alveolar bone resorption: 3D analysis using FEM. Journal of Dentistry, Shiraz University of Medical Sciences 2002:3(3–4):39–49.
11. Geramy A, Faghihi S. Secondary trauma from occlusion: 3D analysis using finite element method. Quintessence Int 2004;35:835–43.
12. Geramy A. the cervical headgear force system: three-dimensional analysis by means of the finite element method. J Dentistry, Shiraz University of Medical Sciences 2000;2(3): 21–30.
13. Geramy A, Morgano S. Finite element analysis of three designs of an implant-supported molar crown. J Prosth Dent 2004;92:434–40.
14. Geramy A. Optimization of unilateral overjet management: three-dimensional analysis by the finite element method the Angle Orthod 2002;72:585–92.
15. Burstone CJ, Koenig HA. Optimizing anterior and canine retraction. Am J Orthod 1976;70:1–19.
16. Raboud DW, Faulkner MG, Lipsett AW, Haberstock DL. Three-dimensional effects in retraction appliance design. Am J Orthod Dentofacial Orthop 1997;112:378–92.
17. Chen J, Markham DL, Kotana TR. Effects of T-loop geometry on its forces and moments. Angle Orthod 2000;70: 48–51.