Research Article

JRAD Journal of Research and Advancement in Dentistry

J Res Adv Dent 2018;7:2:9-17.

A Comparison of Moment-to-Force Characteristics Between off-Centered Preactivated Nickel-Titanium and Titanium-Molybdenum Alloy Symmetrical T-Loops: A Finite Element Study

Reddeppa Reddy Badepalli[1*] Sai Seshadri Bharath Thota[2] Shaik Abdul Khaliq[3] Hallharvi Sriram Reddy[4] Garadappagari Dharmadeep[5] Polisetty Siva Krishna[6]

[1]Senior Lecturer, Department of Orthodontics, G. Pulla Reddy Dental College and Hospital, Kurnool, India.
[2]Senior Lecturer, Department of Orthodontics, Narayana Dental College and Hospital, Nellore, India.
[3]Registrar, NAS Dental Dispensary, Kingom of Saudi.
[4]Consultant Orthodontist, Partha Dental Clinic, Hyderabad, India.
[5]Senior Lecturer, Department of Orthodontics, G. Pulla Reddy Dental College and Hospital, Kurnool, India.
[6]Consultant, Department of Orthodontics, Nellore, India.

ABSTRACT

Aim: To investigate the horizontal forces, vertical forces, moments and M:F ratios generated on anterior and posterior segments during deactivation of closing loops (7 mm - 0 mm) constructed from NiTi and TMA alloys with 3mm off-centering towards beta segment and incorporating preactivation bends of 30^0 on both alpha and beta segments using finite element method.

Materials and methods: To perform 3-dimensiosnal finite element analysis, finite element model of maxilla with MBT bracket system, segmented 0.019 × 0.025 inch SS wires, 0.036 inch diameter transpalatal arch connecting first molars, 0.010 inch SS ligature wire consolidating anterior and posterior segments individually and standard 0.017 × 0.025 inch T-loop retraction systems made of TMA and NiTi were generated by ANSYS software.

Results: Both NiTi and TMA alloy T-loops shows decrease of horizontal and vertical forces, alpha and beta moments during their deactivation ranges, whereas both α and β M:F ratios are increased during deactivation. During some point of deactivation both TMA and NiTi T-loops achieve α M:F ratio $\geq$ 7 and α M:F ratio $\geq$ 10 which can be useful for movement of teeth.

Conclusion: Instead of its low stiffness and flexibility, NiTi T-loops are producing M:F ratios of such type that can be used for control tipping and root torquing during some point of deactivation. So usage of NiTi alloy loops in clinical situations can be applicable after recommending further modification in biomechanics for type of tooth movement needed.

Keywords: Finite element method, Moment -to -force ratio, Nickel-Titanium alloy (NiTi), Off- centered position, Titanium-Molybdenum alloy (TMA), T-loop.

INTRODUCTION

Orthodontists accomplish space closure in different ways, depending on the diagnosis and treatment planning.[1] Frictionless systems of space closure advocated by Bur stone and co-workers[2]are superior to systems which introduce friction as a means of space closure. An appliance without friction allows greater control of tooth movement during space closure. Specialized pre-calibrated

Received: Nov. 11, 2017: **Accepted:** Jan. 2, 2018
***Correspondence Dr. Reddeppa Reddy Badepalli.**
Department of Orthodontics, G. Pulla Reddy Dental College and Hospital, Kurnool, India.
Email: Not Disclosed

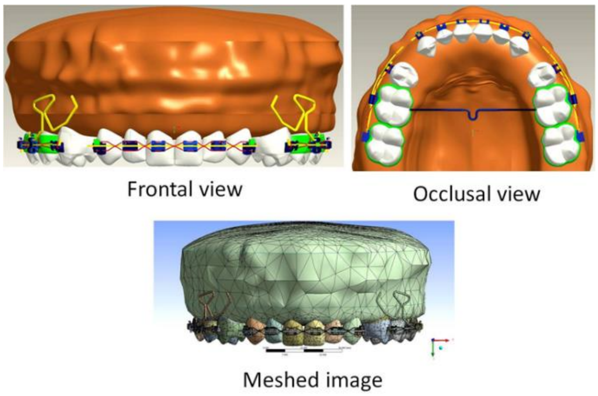

Fig 1: 3D finite element models of the maxilla depicting enmasse retraction of 6 anterior teeth.

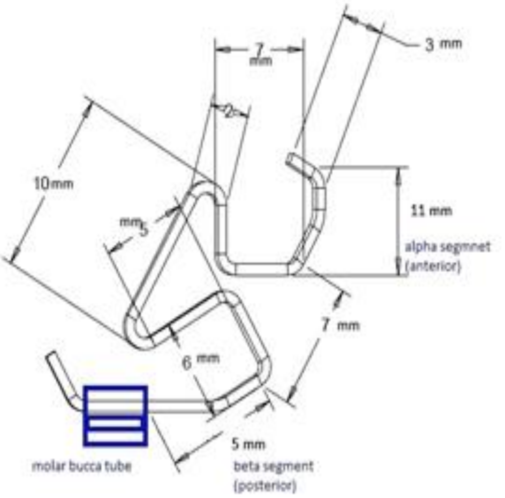

Fig 2: T-loop dimensions: horizontal length -10mm, vertical height of the legs - 6mm(beta) and 7mm(alpha), diameter of the loop - 2mm.

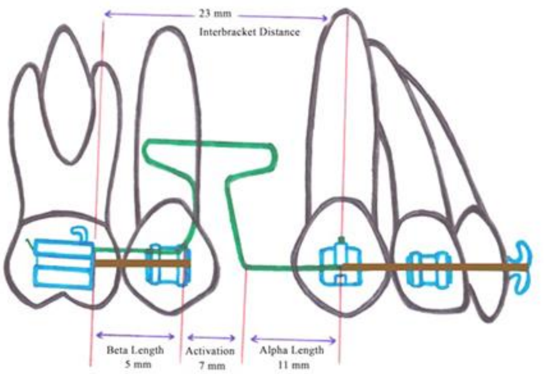

Fig 3: Representing position and activation of T-loop.

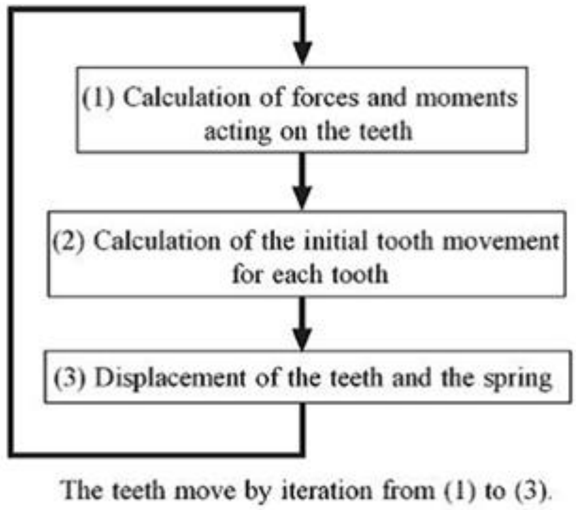

Fig 4: Procedure for simulating the long-term tooth movement.

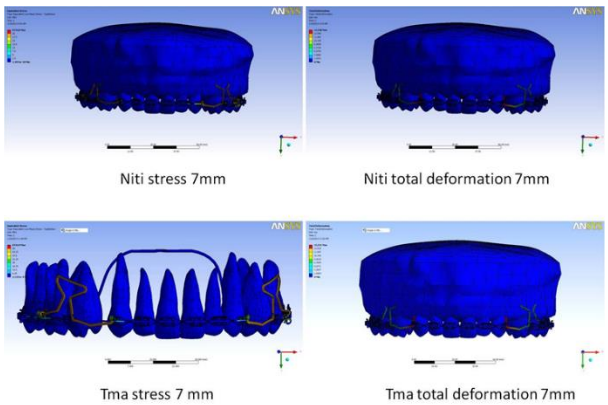

Fig 5: Equivalent Von Misses stresses and Total deformation of TMA and NiTi T-loop retraction systems are represented from respectively during deactivation.

springs for space closure are an integral part of the segmented arch technique. Pre-calibrated attraction springs have three characteristics of interest 1) alpha (anterior segment usually incorporating cuspid and incisor teeth) moment produced by spring 2) beta (posterior segment incorporating molar and bicuspid teeth) moment produced by spring and 3) horizontal force generated. [1] The type of tooth movement is dictated by the moment to force ratio (M/F) generated by the appliance at the attachments. Typically, Moment to Force ratios of approximately 7:1 mm result in controlled tipping; 10:1 mm result in translational movements and values of 12:1 mm (or) greater accomplish root movement. [1] Control of the force system that is

Table 1: Material properties of components of fem model

Material properties	Young's modulus (Mpa)	Poisson's ratio (v)
Tooth	20,000	0.30
Alveolar bone	2,000	0.30
Periodontal ligament (bilinear)	E_1=0.1,E_2=5.0,e_{12}=0.1	0.30
Bracket/Archwire/TPA/Ligature wire made of stainless steel	200,000	0.30
Cortical bone	13,700	0.30
Nickel titanium T-loop	45,500	0.33
Titanium molybdenum T-loop	71,700	0.31

Table 2: Elements and Nodes of structures in FEM Model

STRUCTURES	ELEMENTS	NODES
TEETH	45232	85205
PDL	6619	12781
CORTICAL BONE	22064	42602
CANCELLOUS BONE	25374	46863
BRACKETS	11032	21301
T- LOOP	8826	17041
TOTAL	112898	231836

Table 3: Titanium-Molybdneum Alloy T –Loop Of 3 Mm Off-Centered To Posterior Segment With Symmetrical 30^0 Preactivation Bends.

Deactivation Range	Horizontal Force (gms)	Vertical Force (gms)	Alpha Moment (gms.mm)	Beta Moment (gms.mm)	M/F ratio (alpha)	M/F ratio (beta)
7 mm	448.12	118.44	-2240.6	4570.82	-5.0	10.2
6 mm	385.96	113.17	-2084.18	4168.36	-5.4	10.8
5 mm	323.8	107.8	-1975.18	3658.94	-6.1	11.3
4 mm	261.64	102.63	-1726.82	3165.84	-6.6	12.1
3 mm	199.48	97.36	-1496.1	2792.72	-7.5	14
2 mm	137.32	92.09	-1222.14	2416.83	-8.9	17.6
1 mm	75.16	86.82	-871.85	1991.74	-11.6	26.5
0 mm	-1.7	81.55	-609.7	1425.5	358.7	-838.5

Negative (-) sign indicates clockwise moment direction

Table 4: Nickel-titanium alloy T–loop of 3 mm off- centered to posterior segment with symmetrical 30^0 preactivation bends.

Deactivation Range	Horizontal Force (gms)	Vertical Force (gms)	Alpha Moment (gms.mm)	Beta Moment (gms.mm)	M/F ratio (alpha)	M/F ratio (beta)
7 mm	178.4	55.3	-856.32	1801.84	-4.8	10.1
6 mm	159.3	52.6	-812.43	1688.58	-5.1	10.6

5 mm	143.6	50.4	-804.16	1579.6	-5.6	11
4 mm	125.2	48.3	-801.28	1477.36	-6.4	11.8
3 mm	107.6	46.7	-796.24	1409.56	-7.4	13.1
2 mm	86.3	45.1	-776.7	1311.76	-9.0	15.2
1 mm	41.03	43.6	-594.9	1284.3	-14.5	30.3
0 mm	-0.24	40.8	-70.2	821.16	292.5	-912.4

Negative (-) sign indicates clockwise moment direction.

applied to teeth is one of the main problems in the field of biomechanics. For this reason retraction loops have been used extensively in orthodontics, either in order to produce a force system compatible with desired tooth movement.[3] T-loop introduced by Burstone and co-workers[4] is a simple yet effective device for controlled space closure. The advantages of T-loop design over other loops is that it produces a higher Moment to Force ratio, a lower load deflection rate ratio, delivers a more constant force and Moment to force ratio and helps in avoiding undesirable sequels which include anchorage loss, pain and undermining resorption.[1] Standard form of 0.017x0.025 inch TMA T-loop without pre-activation bends has dimensions of loop length 10 mm, diameter 2 mm, alpha vertical leg height 5 mm and beta vertical leg height 4 mm. Incorporation of pre-activation bends leads to development of moments and M: F ratios.[5]Before application, the spring geometry is adjusted by adding pre-activation bends to the design. The spring is then positioned and activated between the cuspid and the molar attachment tubes (interbracket distance) to produce variations in the delivered force systems. Several of the parameters of the T-loop including the spring height, the placement of the spring within the interbracket distance (clamping position), the amount of pre-activation and axial activation can be changed to appreciably alter the force/moment system delivered to the arch segments.[6] Moment differential can be achieved by off-center positioning of loop and changing the angulation of pre-activation bends. More positioning of loop towards posterior segment produces increased beta moment and more positioning of loop towards anterior segment produces increased alpha moment.[5]Quantitative knowledge of the forces and moments generated by orthodontic springs is crucial for understanding tooth movement. Instruments have been built to measure orthodontic loads. Analytical and computational analyses have been performed to calculate load components.[7] According to the pre-activated bendsthe moment-to-force ratio (M/F) established by the T-loop spring relating to the center of resistance of the tooth and the force can be changed by the orthodontist. This results in better control of the axial movement and makes it possible to select different types of movements.[8]Moment differential obtained by off-center positioning of T-loop remained constant throughout range of deactivation (space closure) reducing the likelihood of anchorage loss.[7]

Initially T-loops were fabricated using stainless steel alloy of specific arch wire dimensions. Later TMA T-loops introduced by Burstone and co-workers [4] replaced stainless steel because of its lower modulus of elasticity than stainless steel. Super elastic NiTi replaced TMA alloys in the fabrication of T-loops because they have low load/deflection rate and low elastic modulus that provide a relatively constant force over a long range of action, can be significantly loaded without permanent deformation and exhibit shape memory.[9] Studies have reported that non-pre-activated NiTi closing loops failed to achieve optimum Moment to force ratio for tooth translation, with an average value of 5 to 5.6.[10]The application of appropriate pre-activation bends might produce a sufficiently high Moment to force ratio so that bodily movement in retracting a single tooth (or) group of teeth (en mass retraction) can occur via translation. The most common pre-activation bends in previous studies of closing loops follow the suggestions of Kuhlberg &Burstone[5]with alpha and beta angles both approximately 30^0 producing a total pre-activation of 60^0. Darnell Rose and Andrew quick[9]showed that centered NiTi loops at 30^0 pre-activation showed a higher mean Moment to force ratio compared with the TMA equivalent and ranged from 10.1 to 39.3g/mm during deactivation of the wire from a 5mm loop opening to a 0.5mm loop opening. Studies by Feng-Yi Keng and Andrew N. Quick[11] identified no difference in the rate of space closure or tooth angulation between pre-activated TMA or NiTi T-loops when all other variables were matched. Finite Element Analysis Method (FEA) is a numerical method used to analyze the stress and deformations occurring in the structure of a geometric model.[12]

The purpose of this study was to investigate the forces, moments and Moment to Force ratios generated on anterior and posterior segments during deactivation of closing loops (7mm- 0mm) constructed from NiTi and TMA alloys with 3mm off-centering and incorporating pre-activation bends of 30^0 on both alpha and beta segments using finite element method.

MATERIAL AND METHODS

CT skull of a 21yrs old male patient with normal anatomy was taken from the records of Department of Oral maxillofacial surgery, G.Pulla Reddy Dental College, Kurnool, India in the axial direction, parallel to the Frankfort horizontal plane. The CT scan images were obtained in a DICOM (digital imaging and communication in medicine) data format. These images were then reconstructed to a 3D model by using Pro/ENGINEER software (Parametric Technology, Needham, Mass). A 3D finite element model of each tooth was constructed according to the method of Wheeler and all teeth were aligned with reference to the facial axis point of Andrews.[12]The labiolingual and buccolingual inclinations of the teeth were simulated based on the studies of Kim et al.The maxillarydentition was established according to the normal arch shapes of Roth (Tru-arch forms, medium size; Ormco, Orange, California). The 3D finite element models of the periodontal ligament were constructed to fit outside the root. The 3D finite element models of the alveolar bone were fabricated to fit the teeth and the periodontal ligament.The finite element method is a highly approved method to simulate biophysical phenomena in computerized models of teeth and their periodontium.[15]ANSYS Workbench 11.0 is a recent version of Ansys which can import models with 100% data transfer or with 0% data loss. Once imported, the software can do an automatic meshing with defined material properties. [Fig 1]

The finite element models were considered to have linear elasticity and isometric properties of the same quality. Young's modulus and Poisson's ratio for the teeth, bracket, and stainless steel wire were calculated according to Reimann and co-workers [13]and PDL according to Gotoand co-workers.[14][Table 1]

Three-dimensional mid-noded tetrahedral solid elements were used for all the structures in the model except PDL and arch wires. The periodontal membrane was constructed with thin shell (triangular surface contact) elements and the arch wires with 3D beam elements. The total number of 112898elements and 231836nodeswere used in the model as shown in [Table 2].

T-loops were modelled with pre-activation bends at gingival level in order to get 7mm separation of vertical legs and incorporating pre-activation bends of 30^0on both alpha and beta segments. The horizontal length of T-loops was 10mm, vertical height of the legs were 6mm(beta) and 7mm(alpha) and diameter of the loop was set at 2mm. Standard interbracket distance was taken as 23 mm [Fig2]. Accurate diagrams of the loops were prepared and the key points marked on each diagram. The key points were used to transfer the exact geometry of each loop to ANSYS software installed in a personal computer. The boundary conditions were defined so that the terminal node in the alpha segment (anterior) was restrained (i.e., 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 posterior segment. Initially the length of the beta segment of loop was marked as 5 mm from the center of loop to mesial end of molar auxiliary tube and length of alpha segment of loop was marked as 11 mm (8 mm+3 mm) from the center of loop to distal end of canine bracket as we are placing the loop 3 mm off-centering and subtracting the activation length (Fig 3). Then the loop was placed into the model, initially beta segment was fixed to the molar bracket. The loop was taken into its neutral position without fixation of alpha segment to canine bracket. Then activation of 7mm was given by placing alpha segment end into canine bracket. Based on study done by Safavi and co-workers [16], insertion of a T-loop with a pre-activation 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 T-loops with preactivation bends.

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 contact between the vertical legs of T-loop produced when the wire was 'engaged'. This point was the neutral position for T- loops (zero force instant) and the starting point for all activations.

5. The different amounts of T- loop deactivation were then applied sequentially and recorded. The output data for the forces, moments, counter moments and M:F ratios of each loop during deactivation (7mm - 0mm) were assessed at the terminal nodes of both alpha and beta segments.

Simulation of long-term tooth movement (Fig 4) was calculated using the same method as done by Kojima and co-workers.[17]

RESULTS

The results involve calculation of stresses by Von Misses criteria and Total deformation for each node. It determines the total state of stress and deformation at a predetermined location. The magnitude of stress and deformation indicates the local sensitivity of the model to the load in question. The areas of stress and total deformation are shown with the help of different colors. The pellets of colors representing the tensile and compressive stresses are shown on the right-hand side of the diagram. Equivalent Von Misses stresses and Total deformation of TMA and NiTi T-loop retraction systems are represented from [Fig 5] respectively during deactivation.

The measured force system for 3 mm off-centered, 30^0 pre-activated symmetrical TMA and NiTi T-loop retraction appliance are shown in [Table 3 and Table 4] respectively. At full activation (7 mm), TMA T-loop system delivered a horizontal force of 448.12 gms whereas NiTi T-loop system delivered less horizontal force of 178.4 gms. In both systems horizontal force decreases during deactivation of loop, but rate of decrease is more in TMA system. Vertical force of 118.44 gms was delivered by TMA T-loop and 55.3 gms was delivered by NiTi T-loop at full activation. In both systems horizontal force decreases during deactivation of loops.

DISCUSSION

3D finite element model (FEM) provides the freedom to stimulate orthodontic force systems applied clinically and allows analysis of the response of the maxillary and mandibular dentition to the orthodontic loads in three-dimensional space.The finite element model is a standard tool used in engineering to precisely assess local stress/strain/displacement distribution in geometrically complex structures. The point of application, magnitude and direction of a force may be easily varied to stimulate the clinical situation.[8]

Off-center positioning maintains the constancy of the moment differential throughout the range of spring deactivation (space closure). This concurs with Burstone and Koenig[7]who demonstrated a moment differential and vertical forces with off-center vertical loops. Faulkner et al[18]evaluated the effects of spring height, pre-activation angulation, spring position, and the addition of helices on the force/moment systems produced by segmented T-loops. Studies by Chang YI and co-workers [19], Sang SJ and co-workers[20], kimand co-workers [21] and Kojimaand co-workers[22]were done using finite element analysis to examine effective en-masse retraction of anterior teeth in different ways.

Based on study done by Kojima and co-workers [17] we have done iteration cycles for long-term movement of teeth during deactivation of TMA and NiTi T-loop retraction systems by 1 mm gradually in the FEM models. In our study we observed that the horizontal forces, alpha and beta moments are decreased during deactivation both for TMA and NiTi T-loops, but rate of decrease is more for TMA alloy loop. So horizontal forces, alpha and beta moments for both TMA [Table 3] and NiTi alloy loops [Table 4] are relatively more than the horizontal forces observed by Darnel Rose and Andrew Quick[9] where centered positioning and symmetrical pre-activation bends of 30^0 were given both for TMA and NiTi T-loops.Vertical forces are relatively increased as observed by Kuhlberg and Burstone[5]where T-loop is centrally placed with symmetrical moments. The moment-to-force ratios increase with deactivation of the springs which resemble the studies of Kulhberg[5] and Rose.[9]

We observed that both TMA and NiTi are producing alpha Moment to Force ratios of >7 during initial deactivation 7 mm to 4 mm leading to uncontrolled tipping of anterior teeth. Later from 3 mm to 2 mm Moment to force ratio<7 and >10 is seen leading controlled tipping. Finally at 1 mm deactivation Moment to force ratio<10 is produced leading to root tipping alone in both alloys. Coming to beta segment for both TMA and NiTi alloy loops,Moment to force ratio<10 is observed initially from starting deactivation range onwards so that root torque can be seen leading to development to good anchorage situation that can be helpful in clinical situations [Table 3 and 4].

One factor that can influence the measurement of a moment is the degree of ‘‘play’’ between the arch wire and the bracket. Here in our study we used the arch wires of same dimensions, whereas studies by Rose and co-workers [9]and Kum and co-workers [10]used different dimensions for TMA and NiTi although they had similar amounts of play, and differed by 0.2^0, which was considered insignificant. By engaging the wires fully into the bracket slots, they justify direct comparisons between the wires without correction for different degrees of engagement into the bracket.

In our study we are using T-loops of same dimension, but accordingly flexibility of NiTi alloy is more when compared to TMA alloy. As a result the forces produced by the NiTi closing springs were generally lower than those produced by the equivalent TMA closing springs for the first two thirds of the deactivation. The springs formed in NiTi alloy then produced a higher mean force, attributed to the hysteresis behavior of NiTi alloy. The deactivation force produced by NiTi alloys is also affected by phase transformations due to either temperature and/or mechanical stress.[23-24]

However, a major disadvantage of using a NiTi alloy is its lack of formability, and the difficulty in placing and adjusting permanent pre-activation bends. A further disadvantage of pre-activated bends is that adjustment is limited to eccentric positioning of the loop between brackets, and as a consequence, a higher moment is produced at the end closer to the loop. This will result in additional vertical forces which may be undesirable. NiTi closing loops delivered a more consistent force than the TMA loops, which although insufficient for enmasseretraction was considered adequate to translate a single tooth.[25-26]

So in our study design of T-loop is not ideal for retraction of anterior segment with the help of alpha moment to force ratio during higher deactivation ranges in spite of its usage in providing good anchorage to posterior segment by beta moment to force ratio from starting of deactivation. Furthermore analytical and computational studies should be recommended with revised and improved biomechanics for making a clinical situation suitable for desired tooth movement without any adverse effects using NiTi and TMA alloys.

CONCLUSIONS

* TMA generally produced a higher mean force over its deactivation range compared with the equivalent NiTi closing-loop specimens.

* All pre-activated TMA and NiTi closing-loop specimens produced a beta Moment to Force>10:1 at some point in their deactivation range, irrespective of the force delivered.

*NiTi T-Loops produced a relatively constant force during deactivation compared to the same design in TMA.

* Usage of NiTi T-loops are recommended in clinical situations where low force system is needed and can be applicable after recommending further modification in biomechanics for type of tooth movement needed.

* T-loop design of our study is not ideal for retraction of anterior segment with the help of alpha Moment to force ratio during higher deactivation ranges in spite of its usage in providing good anchorage to posterior segment by beta Moment to force ratio from starting of deactivation.

CONFLICT OF INTEREST

No potential conflict of interest relevant to this article was reported.

REFERENCES

1. Manhartsberger C, Morton JY, Burstone CJ. Space closure in adult patients using the segmented arch technique. Angle Orthod. 1989; 59:205-10.
2. Burstone CJ. The segmented arch approach to space closure. Am J Orthod. 1982; 82:361-78.
3. Halazonetis DJ. Understanding orthodontic loop pre-activation. Am J Orthod Dentofacial Orthop. 1998; 113:237-41.
4. Burstone CJ. Rationale of the segmented arch. Am J Orthod. 1962; 48:805-22.
5. Kuhlberg AJ, Burstone CJ.T-loop position and anchorage control. Am J Orthod Dentofacial Orthop. 1997; 112:12-8.
6. Chen J, Markham DL, Katona TR. Effects of T-loop geometry on its forces and moments Angle Orthod. 2000; 70:48-51.
7. Burstone CJ, Koenig HA. Optimizing anterior and canine retraction. Am J Orthod.1976; 70:1-19.
8. Singaraju GS, Chembeti D, Mandava P, Reddy VK, Shetty SK, George SA. A comparative study of three types of rapid maxillary expansion devices in surgically assisted maxillary expansion: A finite element study. J Int Oral Health. 2015 Sep; 7:40-6.
9. Rose D, Quick A, Swain M, Herbison P . Moment-to-force characteristics of pre-activated nickel-titanium and titanium-molybdenum alloy symmetrical T-loops. Am J Orthod Dentofacial Orthop. 2009; 135:757-63.
10. Kum M, Quick A, Hood JA, Herbison P.Moment to force ratio characteristics of three Japanese Niti and TMA closing loops. AustOrthod J. 2004; 20:107-14.
11. Keng FY, Quick AN, Swain MV, Herbison P. A comparison of space closure rates between pre-activated nickel–titanium and titanium–molybdenum alloy T-loops: a randomized controlled clinical trial. Eur J Orthod. 2012; 34:33-8.
12. Guven S, Beydemir K, Dundar S, Eratilla V. Evaluation of stress distributions in peri-implant and periodontal bone tissues in 3- and 5-unit tooth and implant-supported fixed zirconia restorations by finite elements analysis. Eur J Dent 2015; 9:329-39.
13. Reimann S, Keilig L, Jäger A, Bourauel C. Biomechanical finite-element investigation of the position of the centre of resistance of the upper incisors.Eur J Orthod. 2007; 29:219-24.
14. Goto T. An experimental study on the physiological mobility of a tooth. ShikwaGakuho. 1971; 71:1415-44.
15. Singh SV, Bhat M, Gupta S, Sharma D, Satija H, Sharma S. Stress distribution of endodontically treated teeth with titanium alloy post and carbon fiber post with different alveolar bone height: A three-dimensional finite element analysis. Eur J Dent 2015; 9:428-32.
16. Safavi MR, Geramy A, Khezri AK. M/F ratios of four different closing loops: 3D analysis using the finite element method (FEM). AustOrthod J. 2006; 22:121-6.
17. Kojima Y, Fukui H. Numerical simulations of canine retraction with T-loop springs based on the updated moment-to-force ratio. Eur J Orthod. 2012; 34:10-8.
18. Faulkner MG, Fuchshuber P, Haberstock D, Mioduchowski. AParametric study of the force/moment systems produced by T-loop retraction springs. J Biomech. 1989; 22:637-47.
19. Chang YI, Shin SJ, Baek SH. Three-dimensional finite element analysis in distal en masse movement of the maxillary dentition with the

multiloop edgewise arch wire. Eur J Orthod. 2004; 26:339-45.

20. Sung SJ, Jang GW, Chun YS, Moon YS. Effective en-masse retraction design with orthodontic mini-implant anchorage: A finite element analysis. Am J Orthod Dentofacial Orthop. 2010; 137:648-57.

21. Kim T, Suh J, Kim N, Lee M. Optimum conditions for parallel translation of maxillary anterior teeth under retraction force determined with the finite element method. Am J Orthod Dentofacial Orthop. 2010; 137:639-47.

22. Kojima Y, Fukui H. Numeric simulations of en-masse space closure with sliding mechanics. Am J Orthod Dentofacial Orthop. 2010; 138:702. -6.

23. Tonner RI, Waters NE. The characteristics of super-elastic Ni-Ti wires in three-point bending. Part I: The effect of temperature. Eur J Orthod. 1994; 16:409-19.

24. Filleul MP, Jordan L. Torsional properties of Ni-Ti and copper Ni-Ti wires: the effect of temperature on physical properties. Eur J Orthod. 1997; 19:637-46.

25. Kuhlberg AJ, Priebe D. Testing the force systems and biomechanics-measured tooth movements from differential moment closing loops. Angle Orthod. 2003; 73:270-80.

26. Melsen B, Fotis V, Burstone CJ. Vertical force considerations in differential space closure. J ClinOrthod. 1990; 24:678-83.