Research Article

The "TA Loop" Better Force Delivery Through Better Design

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ABSTRACT

Introduction: This study evaluates and compares the efficacy of a novel loop design named as TA loop used for anterior teeth retraction by assessing stress distribution and deformation generated on the periodontal ligament, cortical and cancellous bone and compares it with the conventional Mushroom loop using finite element analysis (FEM).

Materials and Method: Finite element models of both the loops and maxilla and mandible was created using ANSYS Software and for geometric modelling, SolidWorks Software was used. The loading condition was designed to mimic retraction forces determined from two different activations and the results were obtained.

Result: FEM affirmed that the stresses generated by the TA loop were within the permissible range which could be taken up by the roots and the surrounding tissues without causing any deleterious effect. The TA loop includes two helices on either side of the horizontal section, resulting in increased wire length and enhanced force delivery.

Conclusion: Significant variations in stress distribution and deformation were observed between the two loops, highlighting the TA loop's superior efficiency in achieving a consistent force delivery due to increased range of actions.

KEYWORDS: Finite element analysis, frictionless mechanics, loops in orthodontics, Mushroom loop, TA (Tarun Anu) loop.

INTRODUCTION

Orthodontic treatment often involves closing extraction spaces, which can be challenging. There are two main methods for space closure: en-masse retraction using sliding mechanics and a dual technique involving canine retraction followed by incisor retraction using loops, which are referred to as frictionless mechanics. The dual technique is advantageous as it places less burden on anchor units as less number of teeth are retracted against a higher anchorage value[1,2]. Loops have been used for a variety of orthodontic movements, ranging from simple alignment to space closure. The use of loops dates back to 1915 when Dr. Ray D Robinson[3] pioneered their usage. In 1933, Dr. Robert Strang[4] incorporated loops into edgewise treatment and highlighted their usefulness in space closure and anchorage control.

Loops have been advocated in orthodontics for various tooth movements by authors such as Tweed[5], Bull[6], Begg[7], Jarabak[8], and Burstone[9]. Materials used for frictionless retraction have also evolved from stiff stainless steel (SS) wires to the more flexible beta titanium wires introduced by Burstone and to the newer materials like Connecticut archwires (CNA) wires which are supposed to reduce the force levels and thus making the treatment more effective and efficient. The Mushroom loop, introduced by Dr. Ravindra Nanda in 2003, is an adaptation of the T-loop and has an archial pattern that reduces the load-deflection rate, producing low and continuous forces. Beta-titanium is recommended for use in the M-loop as it has lower stiffness and promotes constant force delivery[10,11].

The TA loop, named in accordance with its creators, Dr. Tarun Sharma and Dr. Anu Grover, is a unique loop configuration incorporating two helices in the horizontal segment of mushroom loop, which increases the length of wire used. This design is intended to improve the range of action, load/deflection rate, loop recovery and moments produced. Overall, the TA loop is a novel and innovative approach to loop design.

The efficacy of Orthodontic loops is analyzed through experimental and analytical methods, such as finite element method (FEM). FEM is a mathematical model that predicts mechanical phenomena like force, strain, stress, and their interaction with each other. It was originally developed by Turner et al in 1956 and used for the first time, about five decades ago in the aerospace industry to determine the amount of stress on the fuselage. It entered into medical and dental researches in 1970s. It was introduced into Orthodontics by Yettram et al in 1972.[12,13]

Therefore, this study aims to evaluate the efficacy of TA loop over the Mushroom loop by assessing their load deflection rate under two different activations and loop configurations, as well as analyzing stress distribution and deformation in the periodontal ligament and surrounding bone using 3D modeling and finite element analysis (FEA). This study also determines the effects of design stiffness (spring designs), material stiffness (stainless steel and beta titanium) on the force and load deflection rate in order to assess their efficacy and suitability for clinical use.

MATERIALS AND METHOD

Loop Design

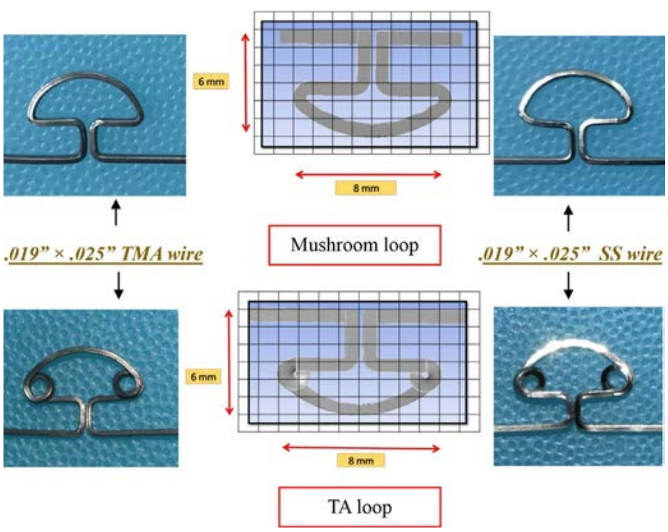


Figure 1:
Dimensions and design of mushroom and TA loop

Both the loops were fabricated using 0.19"× 0.25" TMA wire. Here are some key details -

1. **Dimensions of Mushroom loop –** 6 mm of height, 8 mm of width was taken for the study.
2. **Design of TA loop -** TA loop with the same dimension as that of the Mushroom loop with two helices on each side of horizontal part was fabricated.

Armamentarium used for the formation of loops:

1. β-Titanium wire (.019" × .025")
2. Pliers – no. 142 (Ribbon-arch or tweed), no. 139 (Bird beak)
3. Light wire cutter
4. Permanent marker
5. Glass plate/sim grid

FEM study: methodology

A 3D model of both loops, maxilla and mandible was created using 3D SolidWork software from a cone-beam computed tomography (CBCT) scan of a patientwho meets the study criteria:completion of individual canine retraction completedwith incisor retraction to follow.

Steps for FEM:-

1. Construction of the 3D model.
2. Conversion of the geometric model to a finite element model.
3. Material property data representation.
4. Defining the boundary condition.
5. Loading configuration.
6. Evaluation of stress distribution.

1. Construction of the 3D model

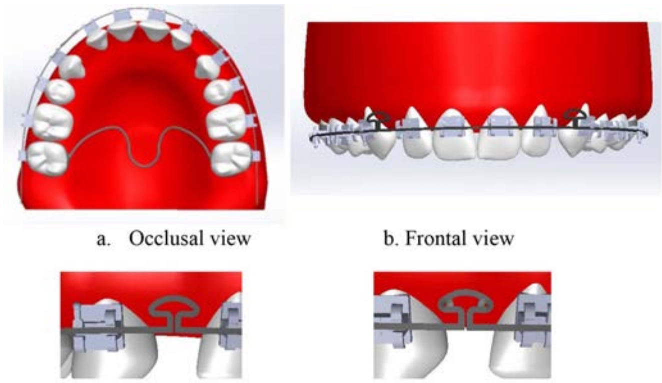


Figure 2:
3D SolidWorks model of maxillary arch with loops

In this study a 3-dimensional CT scan of human maxilla and mandible was taken. The dimensions of teeth for this model were simulated from data obtained through various dental literature. As the thickness of the is not same all over, an average thickness of periodontal ligament was assumed to be 0.25 mm which was generated around the model of the root. The normal apicogingival height of the alveolar bone was considered as 14 mm. 3D model is created using SolidWorks Software from "Dassault Systems SolidWorks Corporation" (Waltham, MA 02451, USA).

2. Conversion of 3D model to finite element model

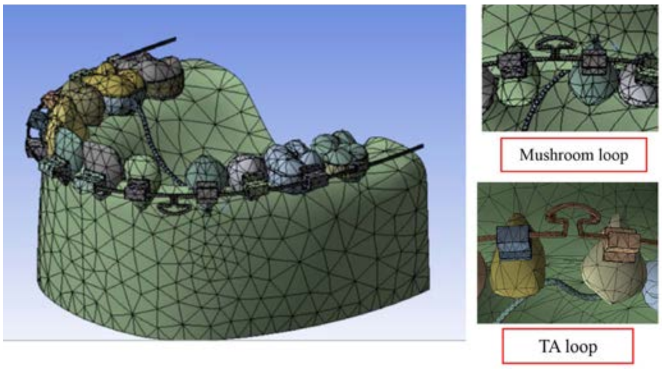


Figure 3: Model meshed in ANSYS

The finite element modelling is the representative of geometry in terms of finite number of element & nodes. This process is called discretization which intends to improve the accuracy of the results. Solid Model is divided into discrete parts called elements which may betetrahedron, rectangular points or hexahedron for 3D analysis.

The ANSYS element library contains more than 100 different element types. We have used solid 186 and Solid 187 elements in this analysis. These elements are considered interconnected at joints which are called nodes or nodal points. The corner nodes are called primary external nodes. The additional nodes which occur on the sides of the elements are called secondary external nodes, which has fewer displacements than corner nodes.

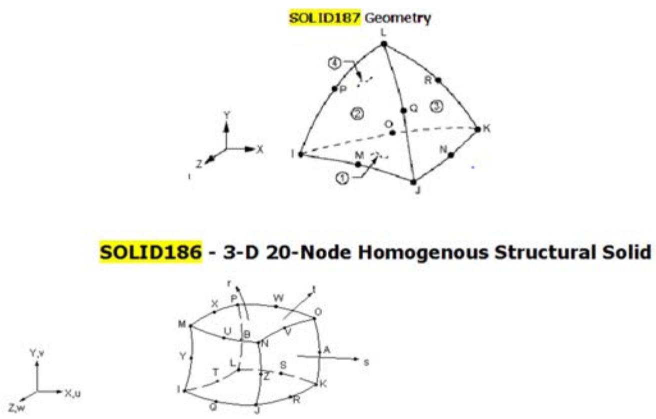


Figure 4 : Elements and nodes

For Finite element analysis, ANSYS Software from ANSYS, Inc. (Canonsburg, PA 15317, USA) is used.

3. Material property data representation:
Material properties of the teeth, periodontal ligament, and surrounding bone were assigned based on previous studies by McGuinness[14] and Tanne K[15]

Table 1: Material properties assigned for FEM analysis

Material	Young's Modulus	Poisson's ratio
Cancellous bone	1340 MPa	0.3
Cortical bone	13400 MPa	0.3
PDL	0.068 MPa	0.45
Dentine	18600 MPa	0.3
Enamel	80000 MPa	0.3
Stainless Steel-brackets	200,000 MPa	0.3
TMA	66 + 1 GPa	0.3

4. Defining the boundary condition

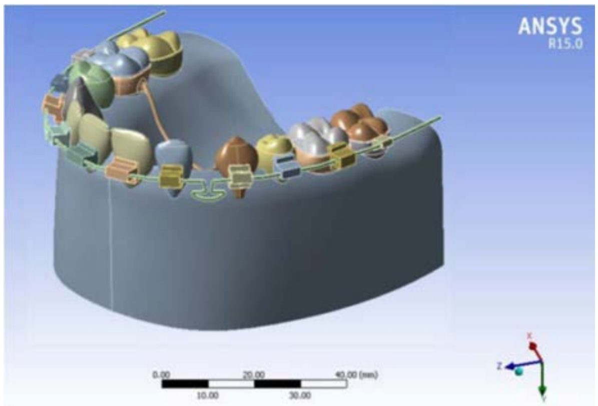
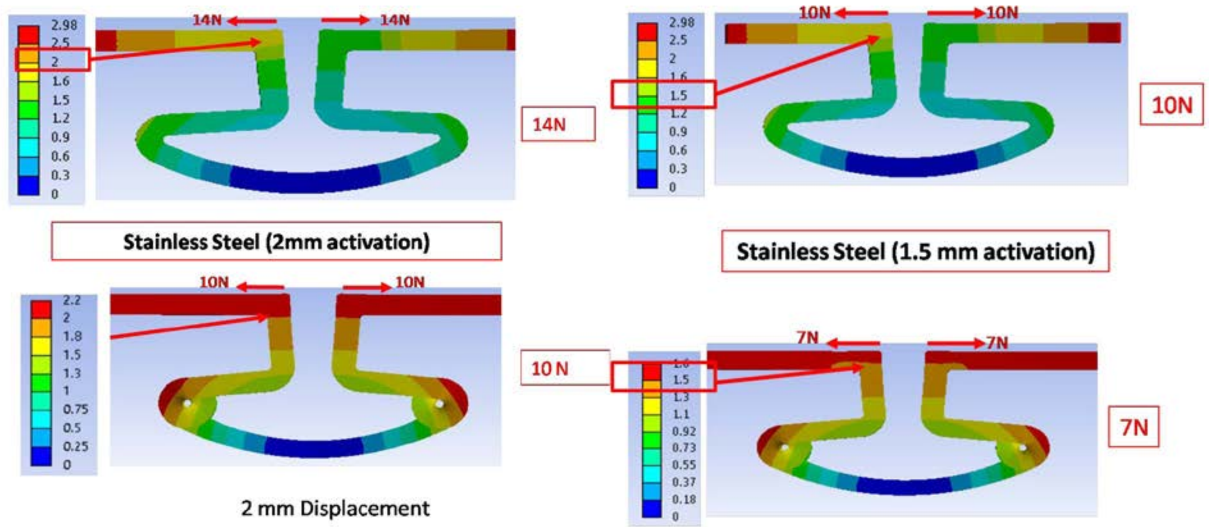


Figure 5 : Fixed boundary conditions

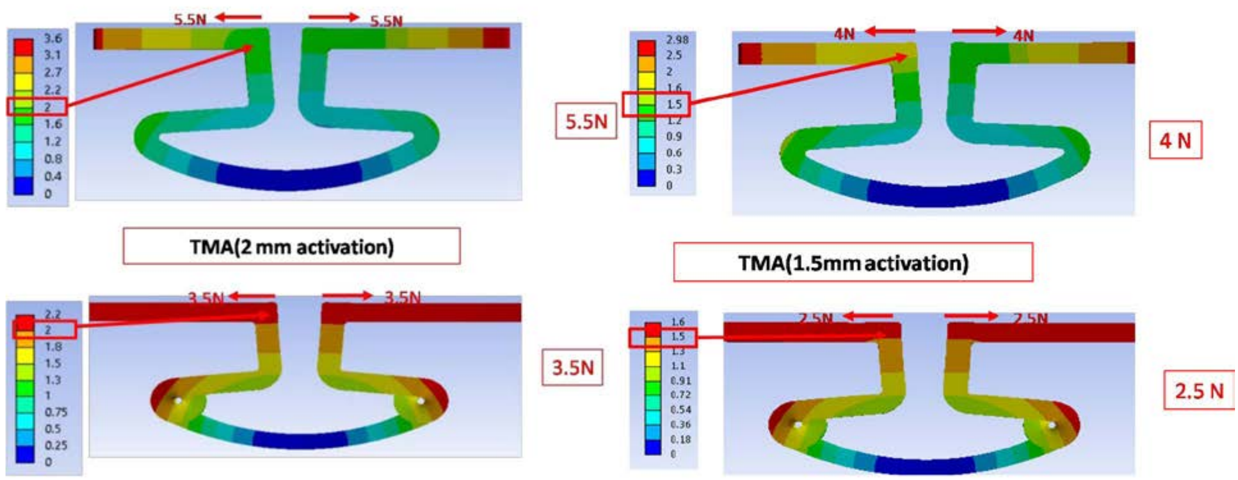
Physical boundary conditions were then applied to the model, including loading, constraints, and other environmental factorsto prevent it from free body motion.

5. Application of force
Force level for two different activations and materials (stainless steel and beta titanium) of both the types of loops were determined using ANSYS software.

The force values obtained from both the loops at 1.5 mm and 2 mm of activation are shown in Table 2



A) .019" × .025" Stainless steel



B) .019" × .025" TMA

Figure 6: Force generatedby Mushroom and TA Loop at two different activations using two different wire materials

Table 2 -Force levels with two different loop designs, wire material and activations

Loop design	Material of wire	Force at 1.5 mm activation	Force at 2 mm activation
Mushroom loop	Stainless steel	10 N	14 N
TA loop	Stainless steel	7 N	10 N
Mushroom loop	Beta titanium	4 N	5.5 N
TA loop	Beta titanium	2.5 N	3.5 N

6. Evaluation of stress distribution.

Stresses (N/mm2) were calculated and presented as different colours, which represented different stress levels. Red colour column of spectrum indicated maximum principal stresses and the following colours like orange, yellow, green and blue represented reducing level of stress with blue colour representing the lowest level of stress. In the study, retraction forces were applied between lateral incisors and canines and evaluation of stress distribution and deformation along the periodontal ligament and the surrounding bone for both loops was carried out.

RESULTS

The results of the study are as follows:-

1. The length of the wire was increased by adding the helices on both sides of the loop which significantly reduced the force levels. There was further reduction in forces when TMA wire was used in comparison with the stainless steel, shown in Fig. 6. Hence, it was found that optimal force levels of approximately 2.5 N (254.9 gms) were obtained with TA loop fabricated from TMA wire at 1.5 mm of activation which was also least when compared with TMA mushroom loop at 1.5 mm activation, stainless steel TA and mushroom loop at 2 mm activation.

2. The stress concentration was found to be more along the lateral incisors followed by canines and conversely, deformation was found to be less along the lateral incisors followed by canines (Fig. 7 and 8). Out of the two loops, Mushroom loop shows higher values of stress distribution and deformation in comparison with TA loop (Table.3)

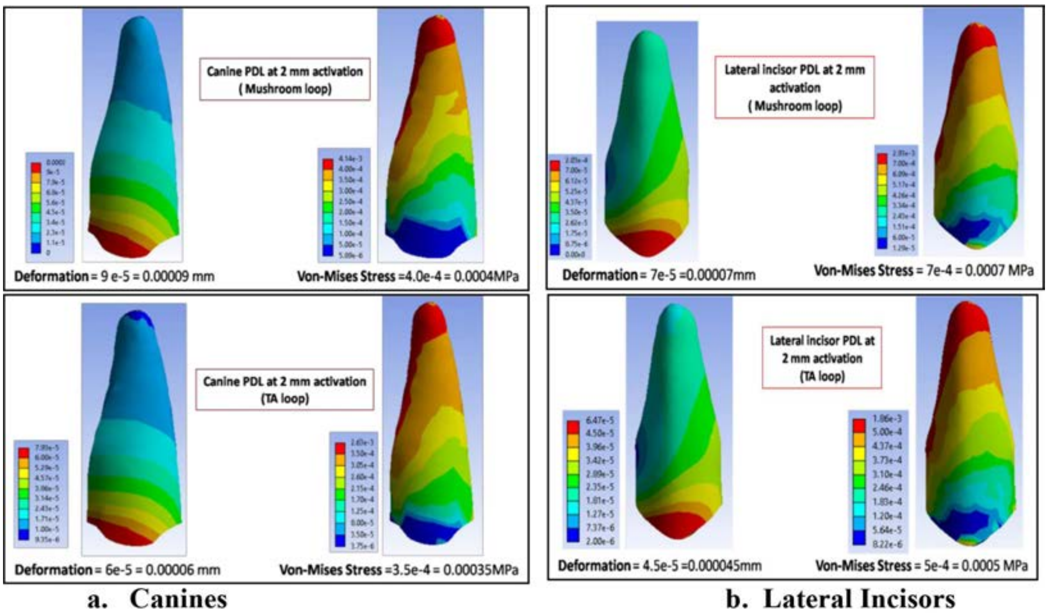


Figure 7: Stress distribution and deformation in periodontal ligament of canines (a) and lateral incisors (b) during retraction of four anterior teeth using mushroom and TA loop at 2 mm of activation.

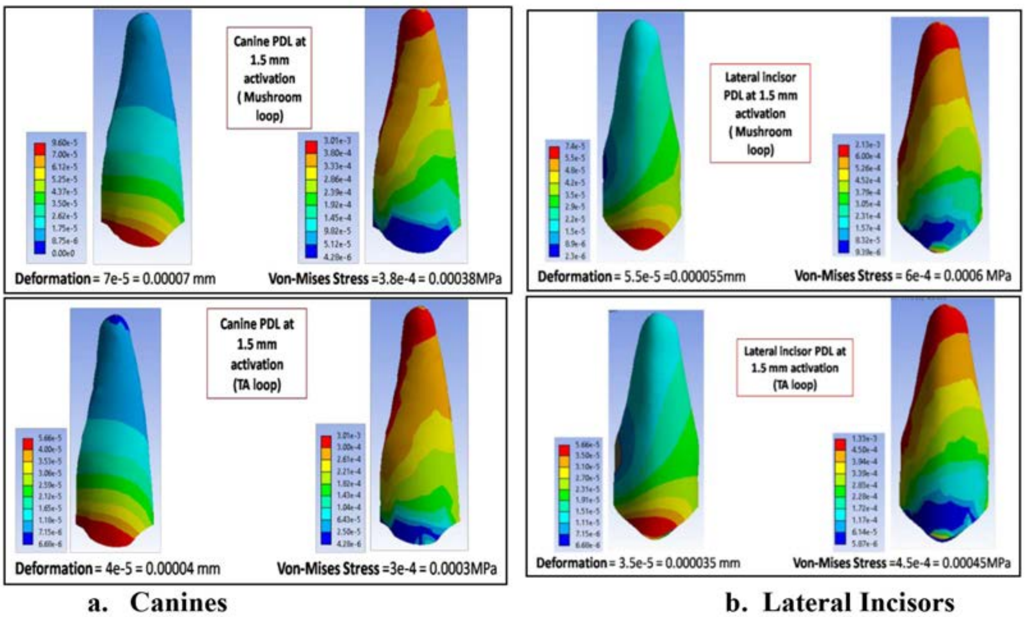


Figure 8: Stress distribution and deformation in periodontal ligament of canines (a) and lateral incisors (b) during retraction of four anterior teeth using mushroom and TA loop at 1.5 mm of activation.

Table 3: Stress distribution and deformation in periodontal ligament of maxillary canine and lateral incisor during retraction of four anterior teeth

Mechanics	Stress distribution and deformation in periodontal ligament of maxillary canine and lateral incisors during retraction of four anterior teeth			
	Deformation (mm)		Von- misses Stress (MPa)	
	Canine	Lateral incisor	Canine	Lateral incisor
M loop 1.5 mm (407.8 grams) 2 mm (560.8 grams)	7×10^{-5} 9×10^{-5}	5.5×10^{-5} 7×10^{-5}	3.8×10^{-4} 4×10^{-4}	6×10^{-4} 7×10^{-4}
TA loop 1.5 mm (254.9 grams) 2 mm (356.8 grams)	4×10^{-5} 6×10^{-5}	3.5×10^{-5} 4.5×10^{-5}	3×10^{-4} 3.5×10^{-4}	4.5×10^{-4} 5×10^{-4}

3. The results showed tensile stress acting in the apex and medial aspect of periodontal ligament of teeth which are more concentrated along the medial aspect of lateral incisors followed by canines while compressive stress acting on cervical margins which are high on lateral incisors than on canines (Fig. 7 and 8)

4. Among the periodontal ligament, cortical bone, and cancellous bone, the cortical bone experienced the highest level of stress compared to the cancellous bone and periodontal ligament. (Fig. 9 and 10). Mushroom loop showed higher maximal stress and deformation on cortical as well as cancellous bone than TA loop comparatively (Table 4 and 5)

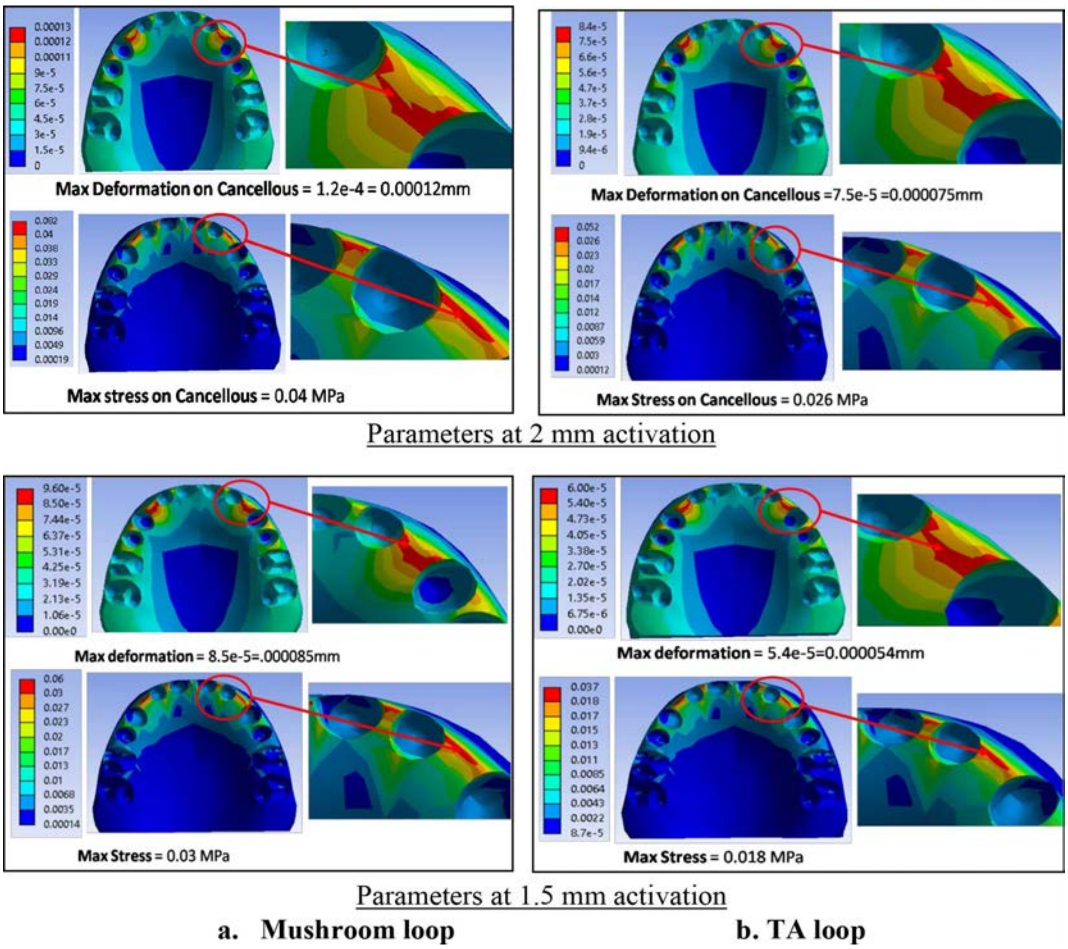


Figure 9: Stress distribution and deformation in cancellous bone by mushroom (a) and TA loop (b) at 2 mm and 1.5 mm) of activation.

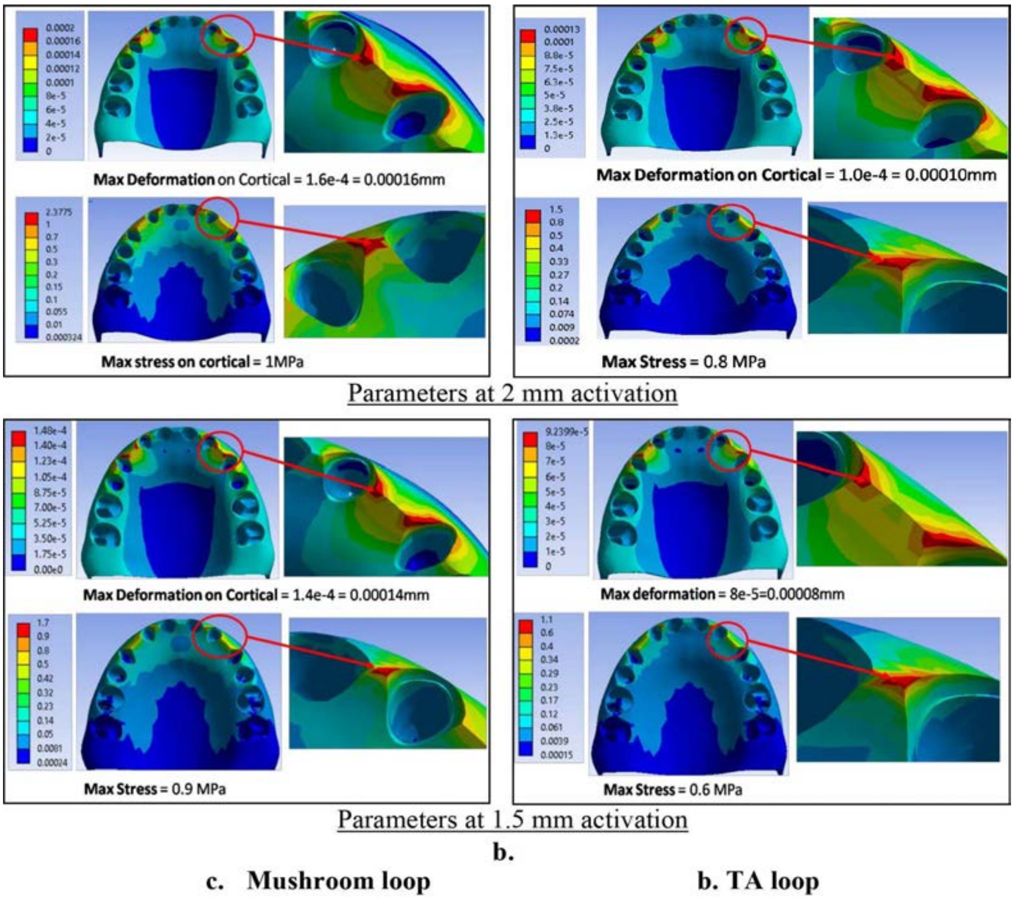


Figure 10: Stress distribution and deformation in cortical bone by mushroom (a) and TA loop (b) at 2 mm and 1.5 mm of activation

Table 4: Stress distribution and deformation in cortical bone of maxillary anterior region during retraction of four anterior teeth

Mechanics	**Stress distribution and deformation in cortical bone of maxillary anterior region during retraction of four anterior teeth**	
	Max. Stress on cortical	**Deformation on cortical**
<u>M loop</u> 1.5 mm (407.8 grams) 2 mm (560.8 grams)	0.9 MPa 1 MPa	1.4 x 10^{-4}mm 1.6 x 10^{-4} mm
<u>TA loop</u> 1.5 mm (254.9 grams) 2 mm (356.8 grams)	0.6 MPa 0.8 MPa	0.8 x 10-4mm 1.0 x 10-4mm

Table 5: Stress distribution and deformation in cancellous bone of maxillary anterior region during retraction of four anterior teeth

Mechanics	**Stress distribution and deformation in cancellous bone of maxillary anterior region during retraction of four anterior teeth**	
	Max Stress on cancellous bone	**Deformation on cancellous bone**
<u>M loop</u> 1.5 mm (407.8 grams) 2 mm (560.8 grams)	0.03 MPa 0.04 MPa	0.85 x 10^{-4}mm 1.2 x 10^{-4} mm
<u>TA loop</u> 1.5 mm (254.9 grams) 2 mm (356.8 grams)	0.018 MPa 0.026 MPa	0.54 x 10^{-4}mm 0.75 x 10^{-4} mm

5. Compressive stress was found to be acting at the cervical margins of both lateral incisors and canines which are more concentrated at cervical margin adjacent to lateral incisors (Fig. 9 and 10). The stress levels were found to be less with TA loop at 2 mm of activation when compared with stress generated with Mushroom loop at the same activation (Table no. 4,5). Similarly, at 1.5 mm of activation TA loop was found to produce less stress levels when compared with Mushroom loop.

DISCUSSION

Loop mechanics has always been the preferred choice for space closure owing to its control in terms of range of action and exhibits the only fail-safe mechanics because force can be contained within the loop for a definitive period of time. The only problem that is associated with loop mechanics are high levels of force which are exhibited because of their rigid setup compared to springs and other force auxiliaries. If we incorporate certain measures in pre-existing designs of loops, it would increase their level of activation, working range and reduce their force levels.

Burstone[16] stated that the overall stiffness of an orthodontic appliance (S) is determined by the wire stiffness (Ws) and design stiffness (As). This relation is presented by the following formula:

$$S = Ws \times As$$

Design stiffness (As) is dependent on factors such as inter-bracket distance and the incorporation of loops and coils into the wire.

Wire stiffness (Ws) can be altered by changing the cross-sectional stiffness (CS) and/or the material stiffness (MS) as designated by the following formula:

$$Ws = MS \times CS$$

The important criteria to be considered for the use of closing loops are given as follows:

Loop position - Off-center positioning of a T-loop produces differential moments. More posterior positioning produces an increased beta moment. More anterior positioning produces an increased alpha moment.[17]

Loop pre-activation - Studies have suggested that the moments occurring through activation alone, arenot sufficient to produce an adequate force system necessary for root control. Recommended beta activation for A, B and C anchorages are 40°, 30° and 20°, respectively. The alpha and beta activation of 20°and 40°, respectively created increased moment on the anchor teeth to preserve anchorage and allow anterior segment to be retracted with adequate root control.[18]

Loop design - Reduction in the force level and increase in moment required for root control can be achieved by increasing the horizontal length of the loop, the height of loop and diameter of bends, or, by adding helices.[18]

Lighter force levels are preferable for dental movement. Ina study, the incorporation of spirals and the use of beta Ti alloys provided lower force levels, lower amounts of horizontal force and load/deflection ratio than stainless steel loops.[6,19]

Keeping all these factors in mind, we designed our TA loop which adds around 12 mm of wire length. Force levels on the two loops in different configurations and subsequent stress on teeth and surrounding bone were determined by using 3D modelling and finite element analysis. The results that came out were very promising as it reduced the force levels to around ideal force levels required by loop.

This study also determines the effects of design stiffness (spring designs), material stiffness (stainless steel and beta titanium) on the force and load deflection rate in order to assess their efficacy and suitability for clinical use using the finite element analysis (FEA).

The TA loop with the same dimension as that of the original Mushroom loop i.e. 6 mm in height and 8 mm in width, with two helices on each side of horizontal part is customised, which increases the overall length of the loop by approximately 6 mm on each side. FEM results showed that there was increase in the magnitude of the force as the displacement increased from 1.5 mm to 2 mm for the two spring designs (Mushroom loop and TA loop) and materials (stainless steel and beta titanium) were also found to influence the magnitude of force. The values were obtained using FEM study and are illustrated Table 2. This is similar to the findings of Geremy[12], Faulkner et al[20], Rodrigues Coimbra[21] and Jadhav et al [22]. The force increased in a linear fashion with each millimetre of activation. This was indicative of the fact that the maximum displacement was under the elastic limit for the loops.

Beta titanium Mushroom loop delivered 0.4 times the force of stainless-steel Mushroom loop. Beta titanium TA loop delivered 0.35 times the force of stainless-steel TA loop. This is similar to Jadhav et al[22], who stated that beta titanium springs delivers 0.38 times the force of stainless-steel springs and findings of Burstone[16] who stated that beta titanium delivers 0.4 times the force of stainless steel.

The Young's modulus taken in the present study was 66 GPa for beta titanium[23] and 200 GPa for stainless steel, therefore, the force exerted by beta titanium was lesser than that of the stainless steel because modulus of elasticity determines the material stiffness, which in turn determines the relative amount of force that a wire can deliver per unit activation.

This difference between the spring designs is attributed to the length of the wire involved in designing the loop. Load deflection rate varies inversely as the cube of the lengths and the change in length can be brought about by either increasing the vertical height of the loop or incorporation of the helices. The vertical height of a particular loop cannot be increased beyond a certain limit due to anatomical considerations. Therefore, addition of helices to the design helps to increase the wire length. On quantifying the load deflection rate of the two loops with and without helices, lower magnitude values and higher constancy for the loops with helices was found. In a study by Allahyar Geremy et al[12], FE analysis showed that M/F was three times greater in a loop with helix in comparison with a loop without helix. The main significant values which were exhibited are as follows- Out of the two loops, mushroom loop shows higher values of stress distribution and deformation in comparison with TA loop. Von misses stress, at 1.5 mm activation, was found to be less along the periodontal ligament of canines and lateral incisors with TA loop i.e., 4.5 x 10-4 MPa along lateral incisor and 3 x 10-4 MPa along the canine, in comparison with Mushroom loop values of 6 x 10-4 MPa along lateral incisor and 3.8 x 10-4 MPa along the canine, suggestive of significant difference between the stress distribution by the loops. Conversely, the deformation was found to be less along the lateral incisors followed by canines.

Out of periodontal ligament, cortical bone and cancellous bone, maximum stress was taken by cortical bone than cancellous bone and periodontal ligament. Mushroom loop showed higher maximal stress and deformation on cortical as well as cancellous bone than TA loop comparatively (Table 4 and 5)

Orthodontics tooth movement is controlled by the elastic response of the PDL through remodeling. Elastic deformation of the PDL determines the initiation of tooth movement. But it was difficult to determine initial tooth movement from present simulation results. This study played a significant role in evaluating the force magnitudes exerted by each loop and their impact on surrounding tissue under varying activation levels and setups. This assessment proved valuable in gauging their potential clinical applicability and effectiveness.

The present study is based on precise evaluation through a predictable, mathematical model which works on the levels of nodes and coordinates. This mimics the precision of artificial intelligence, however, its clinical accuracy is yet to be determined, in-spite of its proven accuracy through finite element modeling. Being a first of a kind approach, clinical studies need to be carried out to obtain real time results of the same.

CONCLUSION

This study helped us in analyzing the stress distribution and deformation of PDL, cortical bone and cancellous bone and most importantly the force exerted by the loop. There was a high level of variance shown in the levels of stress distribution and deformation between the two loops. Therefore, stresses generated by the TA loop (made from .019" × .025" β-Titanium wire) at 1.5 mm activation were within the permissible range which could be taken up by the roots and the surrounding tissues without causing any deleterious effect. Overall increased length of the TA loop by approximately 6 mm on each side led to increase in range of action, facilitating more precise moments and ensured a constant force delivery, making the loop suitable for clinical use.

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