# ANALYSING MECHANICAL PROPERTY OF TWO DIFFERENT CLOSING LOOPS USING FINITE ELEMENT ANALYSIS AND UNIVERSAL TESTING MACHINE - A COMPARITIVE STUDY

*Dissertation submitted to*

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# CERTIFICATE

*This is to certify that the dissertation entitled “**Analysing mechanical property of two different closing loops using finite element analysis and universal testing machine - A comparitive study** done by **Dr. A. Dinesh Kumar**, post graduate student (M.D.S), Orthodontics (branch V), Tamil Nadu Govt. Dental College and Hospital, Chennai, submitted to the Tamil Nadu Dr.M.G.R.Medical University in partial fulfillment for the M.D.S. degree examination (March 2010) is a bonafide research work carried out by him under my supervision and guidance.*

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# DECLARATION

I, Dr.**A.Dinesh kumar** do hereby declare that the dissertation titled "***Analysing mechanical property of two different closing loops using finite element analysis and universal testing machine- A comparitive study***" was done in the Department of Orthodontics, Tamil Nadu Government Dental College & Hospital, Chennai 600 003. I have utilized the facilities provided in the Government Dental College for the study in partial fulfillment of the requirements for the degree of Master of Dental Surgery in the specialty of Orthodontics and Dentofacial Orthopedics (Branch V) during the course period 2007-2010 under the conceptualization and guidance of my dissertation guide, Professor **Dr.M.C.Sainath MDS**. I declare that no part of the dissertation will be utilized for gaining financial assistance for research or other promotions without obtaining prior permission from the Tamil Nadu Government Dental College & Hospital. I also declare that no part of this work will be published either in the print or electronic media except with those who have been actively involved in this dissertation work and I firmly affirm that the right to preserve or publish this work rests solely with the prior permission of the Principal, Tamil Nadu Government Dental College & Hospital, Chennai 600 003, but with the vested right that I shall be cited as the author(s).

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# CONTENTS

## LIST OF TABLES

# LIST OF GRAPHS

# CLOUR PLATES

# INTRODUCTION

Orthodontic biomechanics involves controlled application of moments and forces to bring about desired tooth movement and favourable tissue response. To understand the forces and moments generated on the teeth and surrounding tissues in response to applied load, several experiments were carried out.

The major goals of these experimental testing methods were mainly to predict the optimal force levels that can be applied to the teeth and supporting structures. Among the various in vitro models created to simulate the complex oral environment, finite element analysis used in recent past is the near accurate model to analyse the structural stress /strain relationship of the teeth and surrounding structures in response to the force applied. This method is based on the separation of the analysis shape into sub domains through finite elements. This separation allows a point analysis of the physical behaviour of the object under varied loading conditions.

In fixed orthodontic mechanotherapy, with advent of new orthodontic arch wires, various types of loops were used for space closure. While selecting a loop for orthodontic space closure, a great deal of attention must be paid in terms of the individual clinical problem and anatomical constrains. The common variables that are analysed in selecting loop mechanotherapy include the loop design, its quantity of activation, wire thickness, the metal alloy used, type of movement desired and the amount of force. When choosing loops for closing spaces, it is of utmost importance for the professional to determine precisely the force systems generated, the magnitude of the forces and the moments released when these loops are activated.

Among the various arch wire materials used to construct loops, beta titanium is the best material of choice. β - Titanium wires have improved values of spring back which markedly increases their working range for tooth movement. For a given cross section, it can be deflected approximately twice as far as stainless steel wire without permanent deformation. The high formability of β-titanium allows the fabrication of closing loops with or without helices.

In this study, ‘T’ and Tear drop retraction loops of TMA 0.017 x 0.025 inch rectangular wire were used. Tear drop loops are simple in design with better compliance and flexible making this loop to be most widely used for space closing in both arches. T loop by itself will have a relatively low load-deflection rate and a large maximum spring back. Anchorage can be controlled by varying the location of loop position. The large inter attachment distance between the auxiliary tube on the first molar and the vertical tube of the canine allows sufficient room for the large activations required. Small errors in the shape or geometry of the loop will not radically change the forces.

The purpose of this study was to evaluate the computer simulation and compare it with that of mechanical testing to predict the force obtained from activation of ‘T’ loops and Tear drop loops. ‘T’ loop and Tear drop loop were constructed in FEM with the help of ANSYS software and force were analysed. The Teardrop and 'T' loops were also manually made and force analysed with universal testing machine. The results obtained in both the groups were compared and analysed.

# AIMS AND OBJECTIVES

The aim of this study is to evaluate the force obtained from activation of 'T' loop and Tear drop loop by mechanical testing method and to compare it with force obtained from activation of 'T' loop and Tear drop loop by Finite Element Analysis.

The objectives of the study are:

1. To measure the force exerted by 25'T' loops at various levels of activation by mechanical testing method.
2. To measure the force exerted by 25 Tear drop loops at various levels of activation by mechanical testing method.
3. To measure the force exerted by computer simulated 'T' loop at various levels of activation using Finite Element Analysis.
4. To measure the force exerted by computer simulated Tear drop loop at various levels of activation using Finite Element Analysis.

5. To compare and analyse the force values obtained from mechanical testing method and Finite Element Analysis for 'T' loops
6. To compare and analyse the force values obtained from mechanical testing method and Finite Element Analysis for Tear drop loops.

# REVIEW OF LITERATURE

## BIOMECHANICS

**Schwarz[58] (1932)** detailed the tissue response to the magnitude of the applied force with the capillary bed blood pressure. He concluded that the force delivered as part of orthodontic treatment should not exceed the capillary bed blood pressure (20-26g/cm2 of root surface**).**

**Reitan[51] (1957)** stated that the force exerted in a tipping movement performed with continuous forces will create compressed, cell-free areas in the periodontal membrane more frequently than in a bodily movement, because of the mechanical principles involved. The appropriate amount of force to be applied may vary considerably according to the type of movement required. For example approximately 250 grams of force are required during the final stages of a continuous bodily movement of canines, whereas only 25 grams are required for extrusion of individual front teeth.

**Burstone[10] (1966)** stated that, as an orthodontic appliance is activated, the operator has control over three variables which can determine the success of clinician's adjustment. They are the moment to force ratio, the magnitude of moment or force used, and the constancy of force or moment. He concluded that the segmented mechanism have been designed to control tooth movement which has known moment to force ratio and to aim an optimal biological response by delivery of relatively constant force at an optimal magnitude.

**Weinstein[63] (1967)** explained the efficiency of light forces that is, force of small magnitude in orthodontics. He suggested this by analysing certain areas of the oral musculature and their influence on tooth position. He concluded that there seems to be a genetic factor involved in both the resting force and the stiffness of cheek musculature in males. Masticatory forces produce a buccal tipping tendency on the lower premolar and contribute to the tooth's position of equilibrium.

**Smith and Burstone[9] (1984)** explained about the basic relationships between force and tooth movement and their potential for clinical relevance. Forces produce translation (bodily movement), rotation, or a combination of translation and rotation, depending upon the relationship of the line of action of the force to the center of resistance of the tooth. Since most forces are applied at the bracket, it is necessary to compute equivalent force systems at the center of resistance in order to predict tooth movement.

**Robert. Quinn, Ken Yoshikawa[55] (1985)** stated that the force systems developed by an ideal arch cannot be used directly to estimate tooth movement. They should first be transferred to the centre of resistance of the teeth and the force systems at the centre of resistance may differ significantly from the force systems at the brackets. The relationship between the force magnitude delivered by orthodontic appliances and the rate of orthodontic tooth movement is controversial. He concluded that appliances with low load-deflection rates and relatively constant moment/force ratios allow the clinician to take advantage of the type of tooth movement proposed.

**Birte Melsen, Giorgio Fiorelli. et al [3](1994)**, developed a computer program that can help to design the optimal orthodontic appliance. It involves several phases:

1. Identification of the required tooth movement and its center of rotation.
2. Determination of the force system needed to produce this movement.
3. Appliance design.

With statically indeterminate force systems, calculators are included in the program for complicated calculations to determine forces based on the formulae given by Burstone and Koenig and are available for all possible wire selections.

**Halazonetis[14] (1998)** studied the forces and moments produced by a straight portion of an arch wire which were transferred from the brackets to the center of resistance.

The purpose was to compare the force system at the brackets to the force system at the center of resistance and to assess whether bracket geometry can be applied to predict initial tooth movement. The results show that the force system at the center of resistance may be of an entirely different "geometry" type than that at the bracket. Factors that influence the force system include the interbracket distance, the angulation of the teeth, the length of the tooth root, and the width of the bracket.

**Yijin Ren, DDS, MSca; Jaap C. Maltha[66] (2003)** did a study about the meta-analysis of the literature concerning the optimal force or range of forces for orthodontic tooth movement. Articles on animal experiments were in the majority. Besides variation in species, there was also a wide range of variation in force magnitudes, teeth under study, directions of tooth movement, duration of experimental period, and force reactivation. Furthermore, hardly any experiments were reported that provide information on the relation between the velocity of tooth movement and the magnitude of the applied force. Data from human research on the efficiency of orthodontic tooth movement appeared to be very limited. The large variation in data from current literature made it impossible to perform a meta-analysis. They had concluded that no evidence about the optimal force level in orthodontics could be extracted from literature. Well-controlled clinical studies and more standardized animal experiments in the orthodontic field were required to provide more insight into the relation between the applied force and the rate of tooth movement.

## RETRACTION LOOPS

**Burrstone C.J[7] (1961),** stated that the most constant force at optimal level are derived from spring possessing low deflection rate and high available working loads. His study concluded that Stainless steel have slightly higher maximum load than gold springs of identical rates.

**Burstone and A. Jon Goldberg[8] (1980)** reviewed the gold-based, stainless steel, chrome-cobalt-nickel, and nitinol alloys, as well as beta titanium, a new material for orthodontics. Mechanical properties and manipulative characteristics are summarized to develop a basis for the selection of the proper alloy for a given clinical situation. Beta titanium offers an improvement in the properties of presently designed orthodontic appliances with its increased spring back, reduced force magnitudes, good ductility, and weldability, and also its excellent balance of properties should permit the design of future appliances which deliver superior force systems with simplified configuration.

**Scott R. Drake et al[59] (1982)** the mechanical properties of three sizes of stainless steel (SS), nickel-titanium (NT), and titanium-molybdenum (TM) orthodontic wires were studied in tension, bending, and torsion. The wires (0.016 inch, 0.017 by 0.025 inch, and 0.019 by 0.025 inch) were tested in the as-received condition. The study concluded that a titanium-molybdenum teardrop closing loop delivers less than one half of the force compared to stainless steel loop for similar activations.

**Burstone[6] (1982)** studied the clinical application of frictionless attraction springs using the segmented arch technique .He concluded the most important considerations in the clinical use of attraction springs are the amount of distal activation, the angulation differential between the anterior and posterior teeth, and the centricity or eccentricity of the loop. Loop design lead to a more efficient, hygienic, and comfortable mechanism for space closure.

**Poul Gjessing[45] (1985)** introduced new canine retraction spring in orthodontics .On the basis of a series of theoretical considerations described in the present report, a canine-retraction spring was constructed from 0.016 ´ 0.022 inch stainless steel wire, the principal element being a double ovoid loop 10 mm in height. A "sweep" bend was incorporated to avoid unwanted side effects at the second premolar. He concluded that load deflection and moment/force curves were derived experimentally and demonstrated the ability of the spring to generate and maintain biomechanical conditions necessary.

**Faulkner et al[33]** (1989) studied the effects of several parameters on the force/moment systems produced by ‘T’-loop retraction springs. The springs are studied by using the finite element method and by experimentally measuring the forces and moments from the various designs. The results showed that varying the spring height can produce larger moment to force ratios as the height increases. Changes in the pre activation bends result in asymmetric moment characteristics and also produce large intrusive/extrusive forces. The addition of helices at the bends did little to alter the springs' mechanical characteristics.

**S.J.Chaconas[57] (1989)** studied three contraction arch wires namely double delta spring, contraction torquing arch, and contraction torquing utility arch wire. The activations studied were intrusion, retraction and torque. He stated that if deepening of the bite is indicated, the clinician should use the double delta arch wire which would produce a lingual crown tipping and possibly extrusion of the incisors during retraction. However, if a deep overbite exists prior to anterior tooth consolidation, the contraction torquing or contraction torquing utility arch wires should be used since they were shown to produce the most effective lingual root torquing during incisor retraction. This action would not result in deepening of the bite.

**Thomas R. Katona et al[62]** (1989) review article describes the mechanical properties and clinical applications of stainless steel, cobalt-chromium, nickel-titanium, beta-titanium, and multi stranded wires. Stainless steel wires have remained popular since their introduction to orthodontics because of their formability, biocompatibility and environmental stability, stiffness, resilience, and low cost. Cobalt-chromium (Co---Cr) wires can be manipulated in a softened state and then subjected to heat treatment. Heat treatment of Co---Cr wires results in a wire with properties similar to those of stainless steel. He found Nitinol wires have a good spring back and low stiffness. This alloy, however, has poor formability and weldability. Beta-titanium wires provide a combination of adequate spring back, average stiffness, good formability, and can be welded to auxiliaries.

**Fortschr Kieferorthop. Bourauel C [18] (1994)** , based on the favourable results achieved by the application of pseudoelastic ‘T’ loops in the course of canine retraction, investigated their application to the retraction of maxillary incisors. A modified Burstone ‘T’ loop was made of a pseudoelastic orthodontic nickel titanium wire and then subjected to experimental testing. The clinical application of the pseudoelastic spring was performed using an individualized retraction arch enclosing the whole front segment. His study concluded that the location of the center of resistance of the upper incisors has not been completely clarified. It is thus recommended that this matter should be given further study.

**D. W. Raboud, MSc, et al** [12] (1997) studied the clinical importance of the three-dimensional effects of the force systems supplied by appliance designs used for retraction . His study concluded that the proposed numerical method can accurately determine the force systems resulting from this out-of-plane pre activation as well as the in-plane force systems.

**J. Halazonetis, DDS, MS et al**[15] (1997) tested the accuracy of the computer simulation, various loops that have been evaluated experimentally in the literature were simulated and their response were compared with the results reported previously. In most cases, the results predicted by the computer program were very similar to the experimental values. Results obtained with wire specimens in an experimental apparatus that measures forces and moments, and values obtained from a computer simulation, both contain potential errors and can be different from the values that can be expected in the clinical setting. Development of the program is expected to address some of these problems and increase the validity of the simulation.

**Andrew J. Kuhlberg**[1.] (1997) studied the effect of off-center positioning on the force system produced by segmented 0.017 ′× 0.025-inch TMA 'T'-loops. A 'T'-loop was designed to produce equal and opposite moments in the centered position. The spring was tested in seven positions, centered, 1, 2, and 3 mm toward the anterior attachment, and 1, 2, and 3 mm toward the posterior attachments. The horizontal force, vertical force, and alpha and beta moments were measured over 6 mm of spring activation. The results showed that the alpha/beta moment ratio was dependent only on the spring position, and independent of spring activation. Eccentric positioning of 'T'-loop springs effectively produces a consistent moment differential through the range of spring activation.

**Demetrios J. Halazonetis[16], (1998)** stated that Control of the force system that is applied to teeth is one of the main problems in the field of biomechanics. A major use of loops is in the retraction of canines, where a correct moment to force ratio is essential for bodily movement. The design of the loop influences both the force levels and the moment to force ratio (M/F) in such a way that it is difficult to change the one without adversely affecting the other.

**Marcelo do Amaral Ferreira[35],(1999)** studied the mechanical behaviour of orthodontics closing loops, with three different wire materials (stainless steel, cobalt-chromium and titanium-molybdenum) and with different cross-sections and a double delta design in tension tests. It was hypothesized that loads, after spring activation, and spring rate, are dependent on cross-section, wire material, and activation. The results showed that the loads are dependent on activation, cross-section, and wire material.

Titanium-molybdenum 0.017 × 0.025 inch (Ormco) springs showed the smallest loads and the best spring rate. (b = 84.9 g/mm).

**Jie Chen et al**[28] (2000) studied the effects of T-Loop Geometry on Its Forces and Moments and demonstrated that the moments and forces generated by a T-loop spring are functions of its geometry and gable angle combined with heat treatment. This Study concluded that increasing its vertical or horizontal dimension reduces the load-deflection rate and the moment-to-force ratio. Gable preactivation and stress relieving heat treatment have the opposite effect.

**Kwangchul Choy**[31] **(2002),** made a study on Controlled Space Closure with a Statically Determinate Retraction System. SDRS uses frictionless mechanics, and its statically determinate force delivery system (i.e, magnitude, direction, and point of force application) can be easily established by a single force measurement. He stated that the cantilever spring has a low load-deflection rate; thus, the force produced is relatively constant, and reactivation is often not required. The force direction changes minimally and remains during space closure, as does the axis of rotation. Therefore, unnecessary jiggling of teeth can be minimized.

**Andrew J. Kuhlberg et al[2] (2003)** study was to compare measured tooth movements with the theoretical force system exerted by differential moment closing loops. T-loop springs designed to deliver a differential moment-to-force ratio to the posterior Vs the anterior teeth were used. The anterior teeth, as represented by the canines, were retracted an average of 1.73 mm, whereas the posterior anchorage (molars) moved mesially only 0.50 mm. Furthermore, the canine teeth exhibited tipping or translation, and the molars showed mesial root movement.

**Kum M, Quick A et al[30]** (2004) investigated the loads ("forces"), moments and moment:force ratios (M/F) generated during activation and deactivation of three closing loop designs constructed from two different orthodontic wire alloys. The study concluded that Optimum

M/F ratios for translation are not possible with non-pre activated vertical ‘U’-Loops or symmetrical ‘T’-Loops.

**Thiesen G et al**[61] (2005) did a study to determine the mechanical characteristics of beta-titanium ‘T’-loops with and without helices, with 0 and 180 degree gable bends, constructed from 0.017 inch x 0.025 inch and 0.019 inch x 0.025 inch wire. The horizontal forces and moment/force ratios generated by plain ‘T’-loops with 180 degree gable bends yielded more adequate force systems. He concluded incorporation of helices in the design of ‘T’-loops seems to be unnecessary.

**Renato Parsekian Martinsa** [53]**;** (2008) stated that both curvature and bend pre activated ‘T’ loop produced symmetrical moments with small vertical force. The curvature pre activated ‘T’ loops produced horizontal forces that were lighter than the bend pre activated ‘T’ loops. The curvature pre activated ‘T’ loops produced MF ratios that were approximately 2.5 mm higher than the bend pre activated ‘T’ loops. He concluded the curvature pre activated T loops showed less force decay per 0.5 mm of deactivation (29.8 gf) than the bend pre activated ‘T’ loops.

**Poowadon Koson-ittikul et al[44] (2008)** stated that the LDR(low deflection rate) for one closing loop should be as low as possible so that the force applied to the tooth will be low to minimize pathological effects on the periodontal support and also maintain constant force over a large deflection range. According to the tests, the helical loop with reversed arms provided the lowest LDR among the four simple closing loops designs studied. Wire bending to change the shape and configurations of the loop is the one procedure to improve or change the force/deflection properties of the orthodontics wire material. The major effect of the LDR comes from the amount of the wire incorporated into a closing loop.

**Miceli Beck Guimaraes Blaya et[41] (2009)** analyzed the mechanical behaviour of different orthodontic retraction loops. Two designs of orthodontic loops for closing space were analyzed: teardrop-shaped (T) and circle-shaped loop (C), of two different heights (6 and 8 mm), and two types of orthodontic wires (stainless steel – 0.19’ × 0.25’; TMA – titanium molybdenum alloy – 0.016’ × 0.016’). The group “teardrop-8 mm-TMA” together with the group “circle-8 mm-TMA” presented the lowest mean value, differing statistically from all of the other groups. It was concluded that the alloy of the wire and the height of the loop would be more important than the loop design.

## FINITE ELEMENT ANALYSIS

**Osamu Miyakawa et al[43] (1985)** described a new simulation method to analyze the initial behaviour of the total system comprising orthodontic appliance, teeth, and their supporting structures. It is based on a finite element method which additionally takes account of a rotational degree of freedom. Beam and rod elements are used for finite element idealization of orthodontic appliance. Through spring elements it is connected with the teeth supported by the alveolar structures. The technique of ‘initial strain’ is introduced so as to analyze the effects of a gable bend and activation on the force system which is delivered by the orthodontic appliance. As compared with the photo elastic technique hitherto used, this method serves to investigate systematically and quantitatively the initial aspect of orthodontic tooth movement.

**Raymond E. Siatkowski at al[49] (1997)** introduced new closing loop using FEM (ANSYS soft ware -Swanson Analysis System, Houston,pa) the Opus loop, which is capable of delivering a non varying target M/F within the range of 8.0-9.1 mm inherently, without adding residual moments by twist or bends (commonly gable bends) anywhere in the arch wire or loop before insertion. The resulting precise force systems delivered with non varying M/F can move groups of teeth more accurately to achieve predetermined anteroposterior treatment goals for esthetics and/or stability. The experimental results show that the loops must be bent accurately to achieve their design potential. The negative impact on M/F of various dimensional changes to the loop design is presented. Experimental data is presented illustrating the improved performance of the new design over standard available designs. Suggested applications of the design for varying anchorage requirements are presented, along with a case report in which rigorous protraction requirements were met.

**Young –ll Chang et al [66](2004),** compared the effects of a multi loop edgewise archwire (MEAW) on distal en masse movement with a continuous plain ideal archwire(IA) using finite element method. Three-dimensional finite element models of the maxillary dentition was constructed according to Wheeler and aligned with reference to the facial axis point of Andrews and also standard edgewise bracket and stainless steel IA and MEAW. The stress distribution and displacement of the maxillary dentition were analyzed when classII inter maxillary elastics (300gm/side) were applied. The results showed that the MEAW produced low amount of tooth displacement. Individual tooth movement was more uniform and balanced and less vertical displacement was seen with MEAW.

**Rodrigo F. Viecilli et al**[56](2006) studied about Self-corrective 'T'-loop design for differential space closure .The effects of steps, angles, and vertical forces were combined to produce an ideal 'T'**-loop** design that would provide a more determinate force system. The effects and force systems are estimated based on simplified locations of the centers of resistance, assuming relatively constant behavior of the centers of rotation. These simplifications might differ slightly from what happens in vivo. The study concluded that the **finite element** method or an accurate spring tester capable of reproducing the geometric corrections should be used to ensure a precise force system.

**Mohd. Reza safavi**[42] **(2006**) studied M/F ratio of four different closing loops using FEM. The objective of the study was 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). Results showed that the highest horizontal and vertical forces were produced by the ‘L’–loop (with and without pre activation bends) and in most cases the lowest forces were produced by the VHC loop. He concluded Loops with pre activation bends produced marked changes in the M/F ratio and loops without pre activation bends low, but relatively constant, M/F ratios over the full range of activation.

**Maria Elisa Rodrigues Coimbra**[38] **et al (2008),** studied about Mechanical testing and Finite element analysis of orthodontic teardrop loop. The purpose of this study was to evaluate the use of computer simulation to predict the force and the torsion obtained after the activation of teardrop loops of 3 heights. The computer simulation accurately predicted the experimentally determined mechanical behaviour of teardrop loops of different heights and should be considered an alternative for designing orthodontic appliances before treatment.

# MATERIALS AND METHODS

The present study was conducted to evaluate and compare the force obtained after activation of Tear drop and “T” loops by mechanical method and by finite element method (Computer simulation). Retraction loops of TMA 0.17x 0.25inch rectangular wire were used.

The testing apparatus consisted of computer with ANSYS software and universal testing machine (SHIMADZU Model;AG-IS 50 KN) (Fig 10)

Fifty closing loops made of TMA 0.017 × 0.025-inch rectangular wires were bent. Based on the design of the loops, they were divided into 2 groups namely 25 Tear drop loops[40] (Fig 7) and 25 T loops (Fig 3). Dimension of Tear drop loop was 7 mm height and 2.5mm internal arch circumference. Geometry of T loop was 5mm anterior (α) vertical height and 4mm posterior (β) vertical height. Posterior vertical height was made to be in level with auxiliary tube[37]**.** Width (horizontal length) of loop was

10mm. The same operator prepared the specimens using a template or millimetre jig[41].

A light wire plier with TC tip (Dentronix, *USA*) and Tweed loop forming plier (Rocky Mountain *GERMANY*) were used for making the loops. After preparation, each loop had 28 mm of total length of which anterior and posterior 5mm of wire allowed its attachment to a universal testing machine (Fig 10). Remaining 18mm represented the standard inter bracket distance (IBD)[2] and loops were placed centrally [36].A Digital Vernier Caliper (fig 1) was used to check the dimensions of the loops to maintain a standard dimension (fig 4&8.) Pre activation bends were not given in any loops.

**MATERIALS**

1. 0.017× 0.025 TMA (Omrco, Glendora,CA ) straight wire (Fig 1).
2. Orthodontic light arch plier with TC tip (Dentronix, USA) (Fig 1).
3. Tweed loop forming plier (Rocky Mountain *GERMANY*) (Fig 1).

4. Orthodontic arch wire cutter (heavy duty) (Fig 1).
5. Template or millimeter jig (Fig 2&6). Template prepared with a help of Solid Works software 2005 (DWG-Editor).
6. Digital Vernier Caliper (Mitutoya, Japan) (Fig 1).
7. A Personal computer with the following configuration was used:

   Monitor -- IBM TFT Monitor

   CPU -- IBM (Intellistation Z Pro)

   Processor -- Intel Xeon (Dual Processor)

   Memory Capacity -- Primary -2 GB, Secondary – 80 GB

   Graphics card -- ATI FireGL V 71

   Software -- ANSYS (version 11,)

   Operative system -- Windows vista

## METHODOLOGY:

**Evaluation with Universal testing machine.**

The springs were then subjected to a tensile load on the universal testing machine. (SHIMADZU model: AG-IS 50KN, *JAPAN*). A load cell of 20.0 N was used[36]. For this purpose, one end of the specimen was fixed to the machine, and the other end was displaced. The loops were subjected to activation steps with increments of 0.5 mm at a rate of 1.0 mm per minute up to a maximum displacement of 2mm. The forces were measured and recorded.

**Evaluation using Computer simulation:**

The geometry of the springs used in the computer simulation was obtained with the help of jig dimension. The Tear drop loop (Fig 9) and ‘T’ loop models (Fig 5) were created in ANSYS software according to the characteristics of the loop structures and also considering the specific movements imposed by the intended mechanical activation. BEAM 4 elements were used for constructing loop models for analysis[36]. This (finite) uni axial element could respond to tension, compression, traction, and torsion movements.

The BEAM 4 element has 6 degree of freedom, 3 translation and 3 rotation points around the axis[36].The modulus of elasticity of the orthodontic wire used in the FEA analysis of the springs was 71.70GPa with a Poisson coefficient of 0.130 (**Siatkowski,**AJO 1997)[48]. Both the Tear drop loop and the 'T' loop models were discretized into finite elements, and an average of 237 elements were used for modelling. To simulate the activation similar to universal testing machine, 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)[41,36]. 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. Force exerted on the horizontal axis during every 0.5mm incremental activation of the loop was determined up to a maximum of 2mm activation (Fig 14 & 15). Both the Tear drop loop model and the 'T' loop model were subjected to this simulated activation and the forces exerted by these loops were recorded.

***FIG 1- ARMAMENTARIUM***

***FIG 2 - TEMPLATE OR MILLIMETER JIG FOR T LOOP***

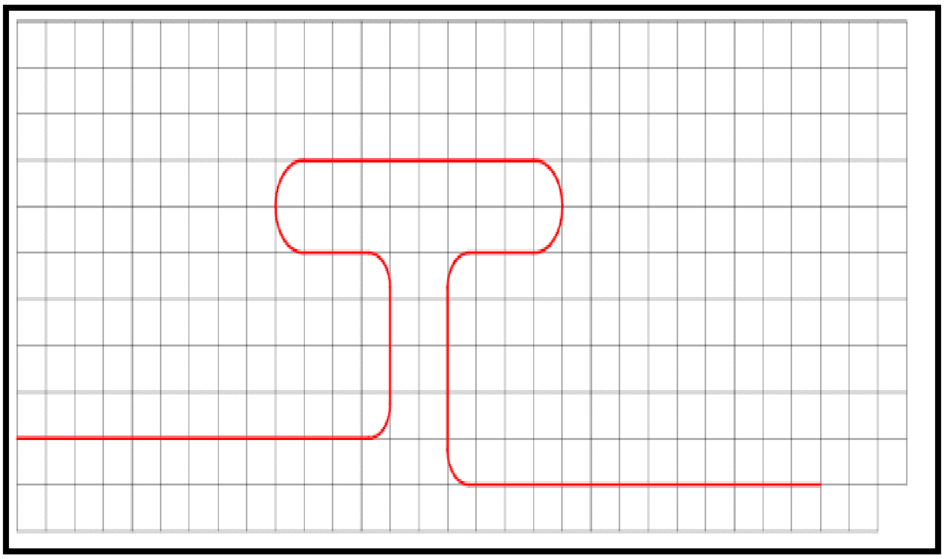

***FIG 3- T LOOP PLACED ON THE JIG MANUALLY MADE***

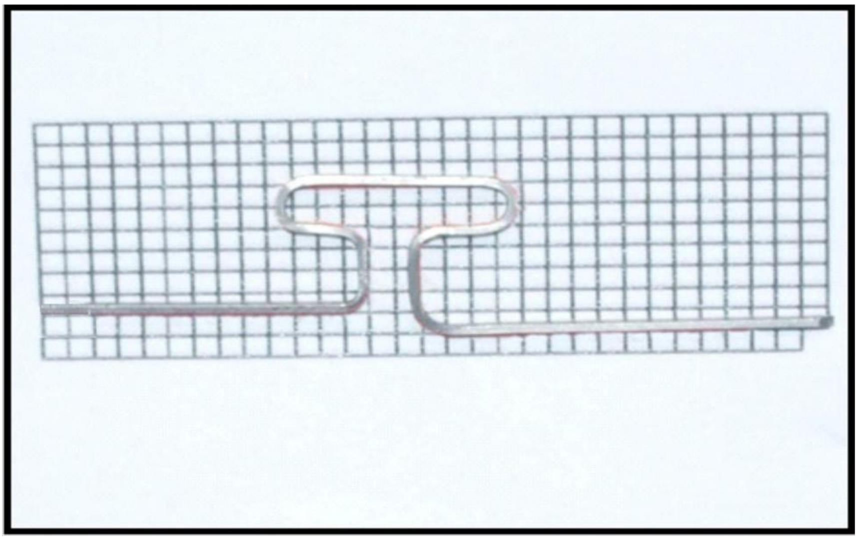

***FIG 4 - MEASURING T LOOP DIMENSION USING DIGITAL CALIPER***

*FIG 5 - MODELING AND MESHING OF T LOOP (COMPUTER SIMULATION)*

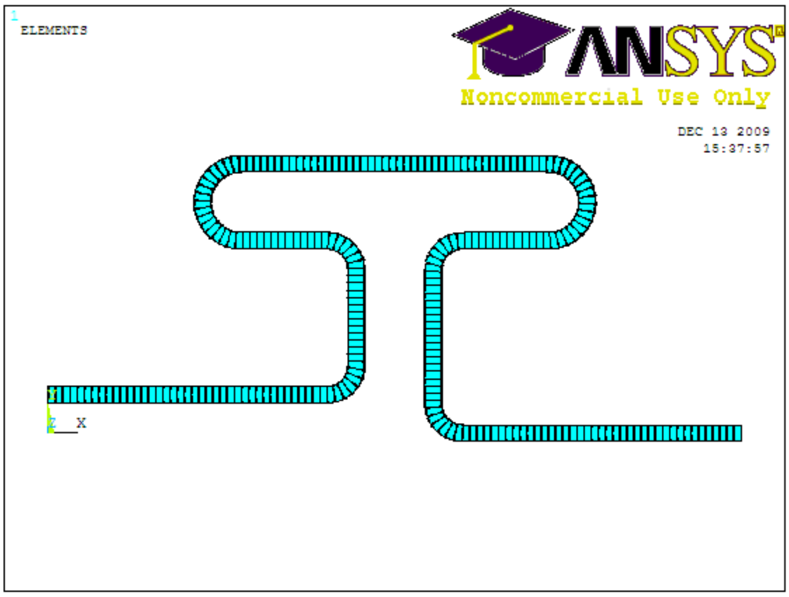


*FIG 6- TEMPLATE OR MILLIMETER JIG FOR TEAR DROP LOOP*

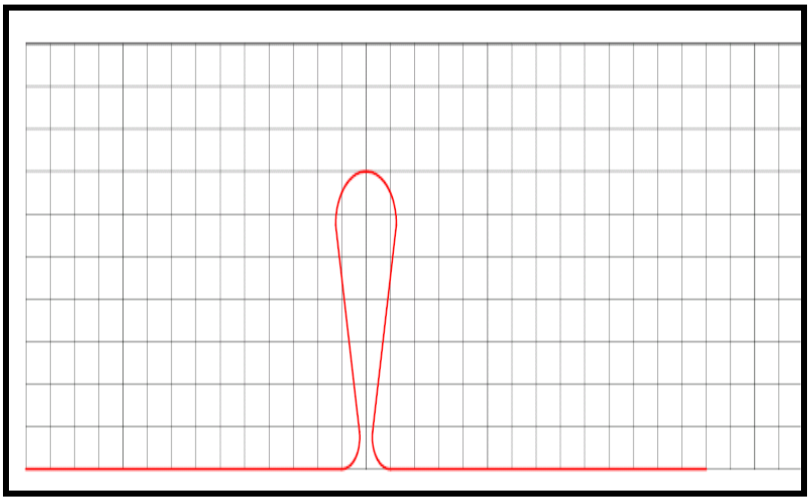

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*FIG 7- TEAR DROP LOOP PLACED ON THE JIG MANUALLY MADE*

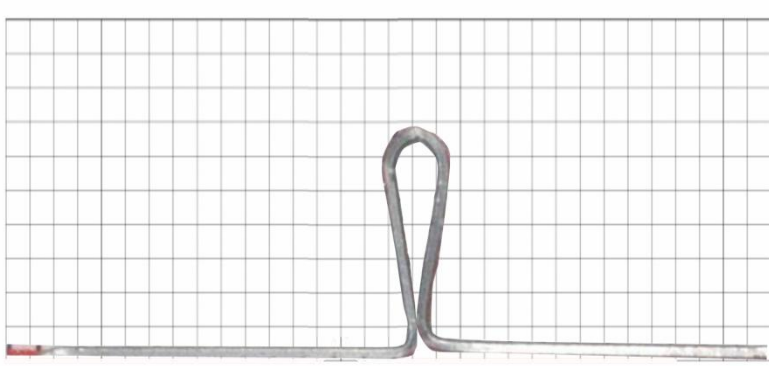


*FIG 8 - MEASURING TEAR DROP LOOP DIMENSION WITH DIGITAL CALIPER*


*FIG 9 - MODELING AND MESHING OF TEAR DROP LOOP (COMPUTER SIMULATION)*

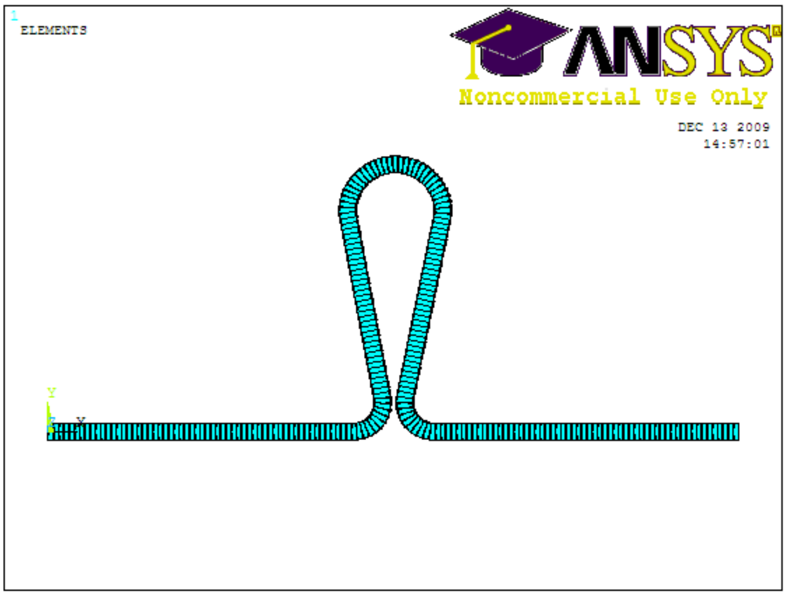



*FIG   10 - UNIVERSAL TESTING MACHINE*

*FIG  11 - T LOOP BEFORE ACTIVATION*

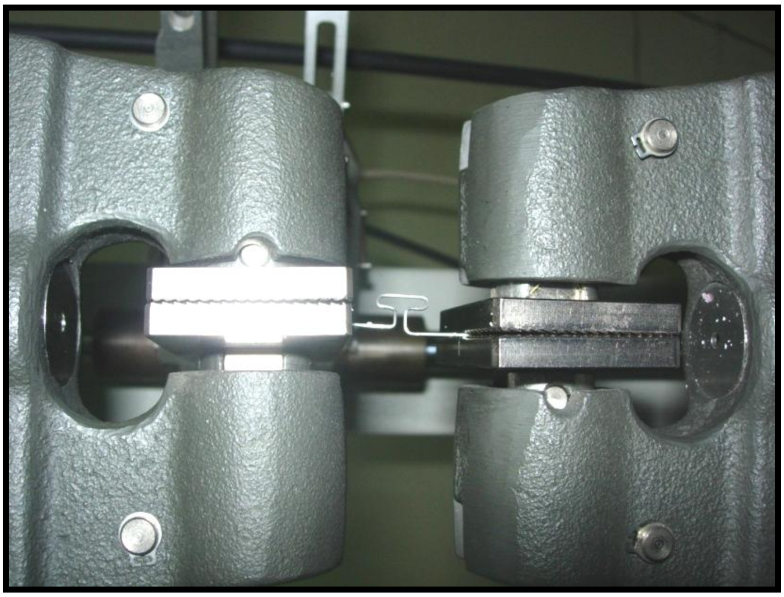


*FIG 12 - TEAR DROP LOOP BEFORE ACTIVATION*

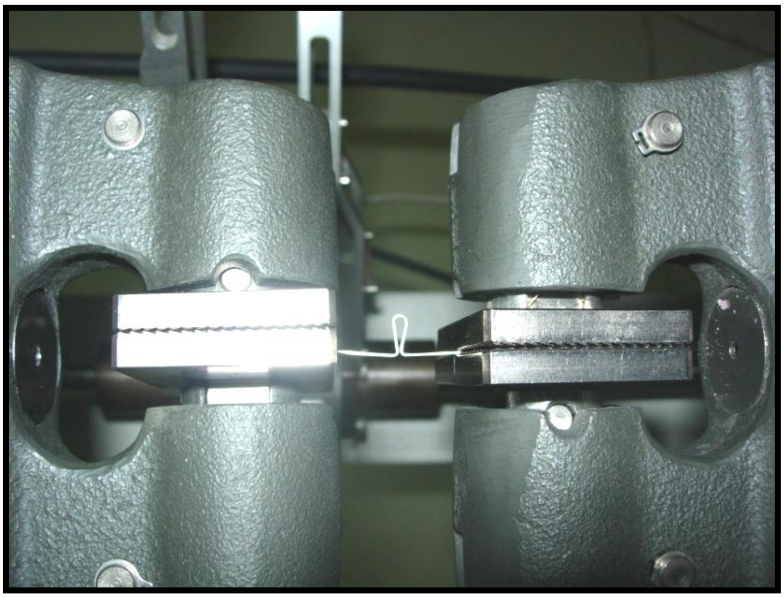


*FIG 13- T LOOP AFTER ACTIVATION*

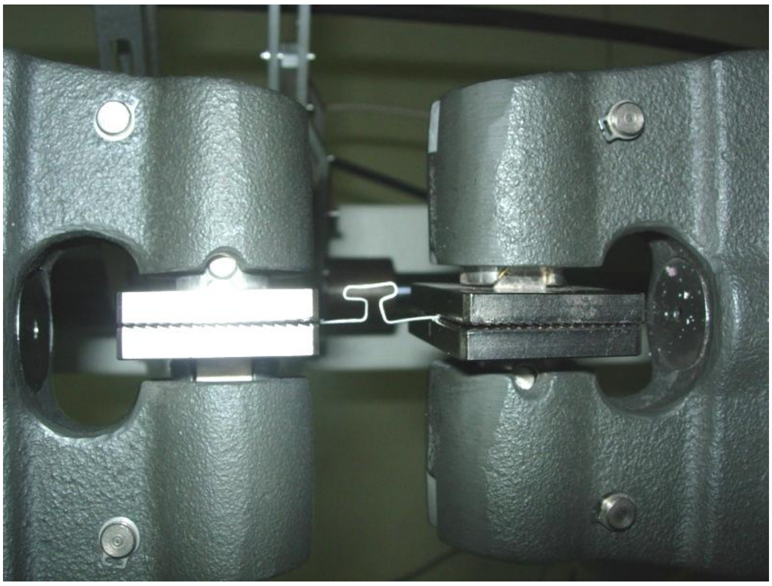


*FIG 14 - 2MM ACTIVATION OF T LOOP (COMPUTER SIMULATION)*

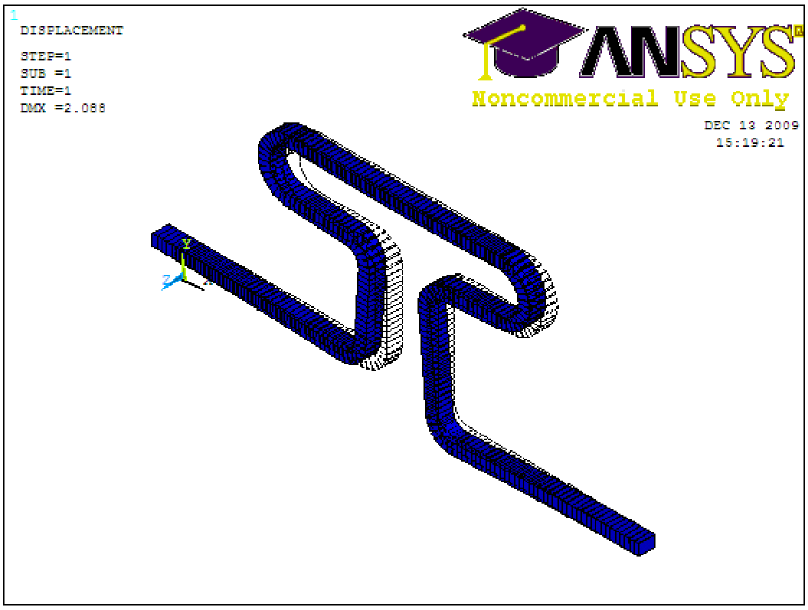



*FIG 15 - TEAR DROP LOOP AFTER ACTIVATION*

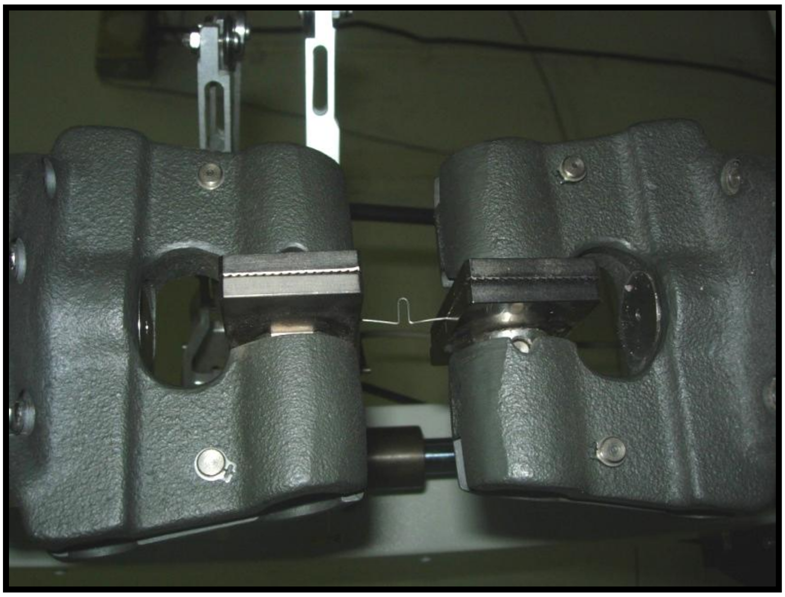


*FIG 16 - 2MM ACTIVATION OF TEAR DROP LOOP (COMPUTER SIMULATION)*

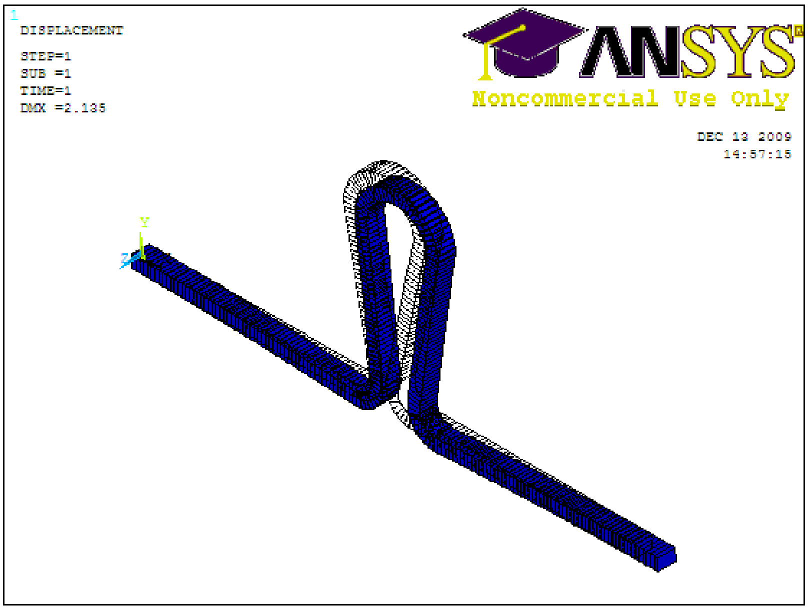



*FIG 17- VONMISES STRESS DURING 2MM ACTIVATION OF T LOOP*

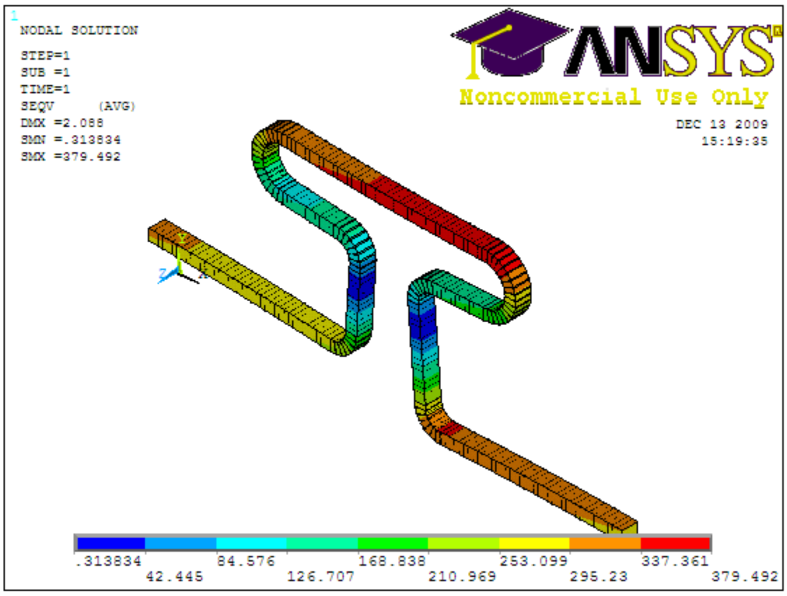
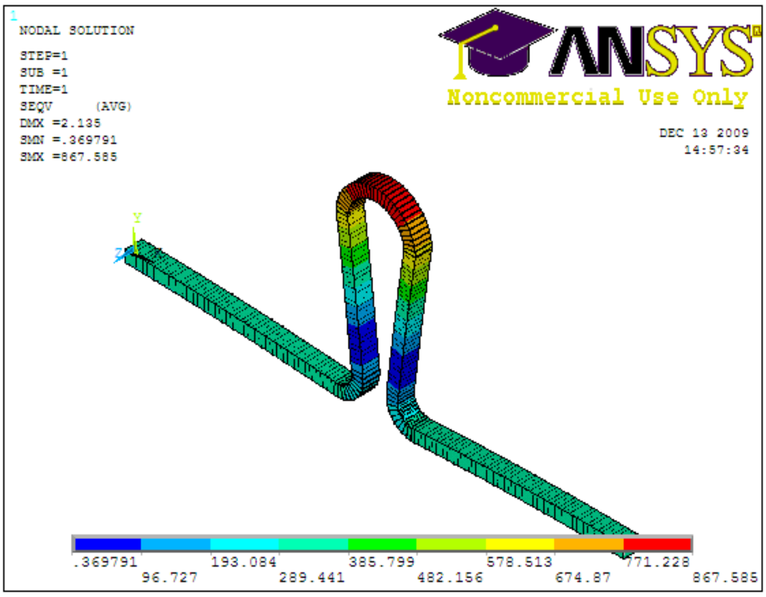



*FIG 18- VONMISES STRESS DURING 2MM ACTIVATION OF TEAR DROP LOOP*



# RESULTS

1. Force value at various levels of activation (from 0.5 mm to 2 mm) for ‘T’ loop and Tear drop loop of the 25 samples each were obtained from mechanical testing. The results are presented in Table 1 and 2 respectively.
2. Computer simulated force value of ‘T’ loop and Tear drop loop were obtained and presented in Table 3 and 4 respectively.
3. Unpaired ‘t’ test showed statistically significant difference between T loop and Tear drop loop force values. The data is presented in table 5.
4. Paired ‘t’ test analysis for computer simulation and experimental testing of ‘T’ loop and Tear drop loop were presented in table 6 and table 7 respectively.

# TABLES

**Table 1.** Force values obtained from T loops at various levels of activation in universal testing machine (Newton).

| T loop Sample no | **0.5mm Activation** | **1mm Activation** | **1.5mm Activation** | **2mm Activation** |
|---|---|---|---|---|
| 1 | 0.53 | 1.024 | 1.546 | 2.165 |
| 2 | 0.403 | 1.085 | 1.446 | 2.435 |
| 3 | 0.512 | 0.987 | 1.418 | 2.45 |
| 4 | 0.516 | 0.906 | 1.616 | 2.21 |
| 5 | 0.501 | 0.916 | 1.483 | 2.333 |
| 6 | 0.511 | 0.953 | 1.439 | 1.904 |
| 7 | 0.416 | 1.09 | 1.442 | 1.867 |
| 8 | 0.523 | 0.952 | 1.341 | 1.849 |
| 9 | 0.609 | 0.932 | 1.372 | 1.911 |
| 10 | 0.458 | 0.963 | 1.392 | 1.891 |
| 11 | 0.556 | 0.961 | 1.462 | 1.54 |
| 12 | 0.512 | 0.966 | 1.468 | 2.354 |
| 13 | 0.534 | 0.956 | 1.429 | 1.825 |
| 14 | 0.509 | 0.903 | 1.421 | 1.834 |
| 15 | 0.514 | 0.912 | 1.425 | 1.815 |
| 16 | 0.52 | 0.916 | 1.556 | 2.435 |
| 17 | 0.59 | 0.953 | 1.486 | 2.45 |
| 18 | 0.454 | 1.09 | 1.468 | 2.21 |
| 19 | 0.553 | 0.961 | 1.586 | 2.333 |
| 20 | 0.619 | 0.928 | 1.462 | 1.954 |
| 21 | 0.459 | 0.953 | 1.466 | 1.967 |
| 22 | 0.566 | 0.963 | 1.458 | 1.849 |
| 23 | 0.512 | 0.956 | 1.437 | 1.891 |
| 24 | 0.534 | 0.957 | 1.432 | 1.54 |
| 25 | 0.509 | 0.913 | 1.473 | 2.454 |
| Average | 0.507 | 0.967 | 1.448 | 2.026 |

**Table 2.** Force values obtained from Tear drop loops at various levels of activation in universal testing machine (Newton).

| | **0.5** | **1** | **1.5** | **2** |
|---|---|---|---|---|
| 1 | 0.995 | 1.76 | 2.78 | 3.24 |
| 2 | 0.831 | 1.47 | 1.92 | 3.6 |
| 3 | 0.841 | 1.545 | 2.359 | 3.227 |
| 4 | 0.842 | 1.86 | 2.743 | 3.515 |
| 5 | 0.825 | 2.08 | 2.99 | 3.49 |
| 6 | 0.805 | 1.77 | 2.12 | 3.306 |
| 7 | 0.902 | 1.436 | 1.859 | 3.617 |
| 8 | 0.934 | 1.352 | 2.001 | 3.054 |
| 9 | 0.801 | 1.647 | 2.125 | 3.232 |
| 10 | 0.729 | 1.551 | 2.75 | 3.251 |
| 11 | 0.847 | 1.391 | 2.05 | 3.598 |
| 12 | 0.742 | 1.669 | 2.078 | 3.263 |
| 13 | 0.745 | 1.496 | 2.161 | 3.004 |
| 14 | 0.732 | 1.379 | 2.005 | 3.058 |
| 15 | 0.807 | 1.352 | 1.974 | 3.021 |
| 16 | 0.835 | 1.536 | 2.752 | 3.543 |
| 17 | 0.835 | 1.353 | 2.152 | 3.317 |
| 18 | 0.85 | 1.648 | 2.118 | 3.582 |
| 19 | 0.827 | 1.553 | 2.151 | 3.091 |
| 20 | 0.815 | 1.431 | 2.015 | 3.242 |
| 21 | 0.892 | 1.659 | 1.963 | 3.241 |
| 22 | 0.934 | 1.486 | 2.121 | 3.599 |
| 23 | 0.728 | 1.369 | 1.959 | 3.252 |
| 24 | 0.837 | 1.472 | 2.091 | 3.015 |
| 25 | 0.741 | 1.535 | 2.152 | 3.148 |
| Average | 0.827 | 1.584 | 2.261 | 3.298 |

**Table 3.** Descriptive statistics for the force level (in N) related during varying activation levels for 'T' loop

| Activation(mm) | Mean (N) | Maximum(N) | Minimum(N) | SD |
|---|---|---|---|---|
| 0.5 | 0.507 | 0.619 | 0.403 | 0.05 |
| 1.0 | 0.967 | 1.09 | 0.903 | 0.053 |
| 1.5 | 1.448 | 1.616 | 1.341 | 0.06 |
| 2.0 | 2.026 | 2.454 | 1.454 | 0.2 |

**Table 4**. Descriptive statistics for the force level (in N) related during varying activation levels for Tear drop loop.

| Activation (mm) | Mean(N) | Maximum(N) | Minimum(N) | SD |
|---|---|---|---|---|
| 0.5 | 0.827 | 0.995 | 0.728 | 0.068 |
| 1.0 | 1.584 | 2.08 | 1.352 | 0.208 |
| 1.5 | 2.261 | 2.08 | 1.859 | 0.342 |
| 2.0 | 3.298 | 3.617 | 3.004 | 0.322 |

**Table 5.** Computer simulation results for T loop

| **Activation** ( mm) | **Force** in Fx (N) |
|---|---|
| 0.5 | 0.481 |
| 1.0 | 0.960 |
| 1.5 | 1.4416 |
| 2.0 | 1.920 |

*x- Force in horizontal axis, N- Newton,*

**Table 6.** Computer simulation results for Tear drop loop

| **Activation** (mm) | **Force** in Fx (N) |
|---|---|
| 0.5 | 0.809 |
| 1.0 | 1.618 |
| 1.5 | 2.427 |
| 2.0 | 3.236 |

*x- Force in horizontal axis, N- Newton,*

**Table 7.** Tear Drop and T Loop Unpaired Students T test

**0.5mm** activation

| Loop | Mean | SD | P |
|---|---|---|---|
| T loop | 0.5168 | 0.05193 | |
| Teardrop loop | 0.826 | 0.06893 | < 0.001 |

**1mm** activation

| Loop | Mean | SD | P |
|---|---|---|---|
| T loop | 0.96384 | 0.05422 | |
| Teardrop loop | 1.552 | 0.17866 | < 0.001 |

**1.5mm** activation

| Loop | Mean | SD | P |
|---|---|---|---|
| T loop | 1.460 | 0.060 | |
| Teardrop loop | 2.245 | 0.0318 | < 0.001 |

**2mm** activation

| Loop | Mean | SD | P |
|---|---|---|---|
| T loop | 2.05 | 0.28 | |
| Teardrop loop | 3.300 | 0.20 | < 0.001 |

*This difference is considered to be statistically highly significant.*

**Table 8**. Paired **'t'** test for T loop between computer stimulation and experiment.

| Activation (mm) | Experimental valve for T loops (N) | Computer simulation value(N) |
|---|---|---|
| 0.5 | 0.507 | 0.48 |
| 1.0 | 0.967 | 0.96 |
| 1.5 | 1.448 | 1.44 |
| 2.0 | 2.026 | 1.92 |

*P value = 0.1915 (not statistically significant.)*

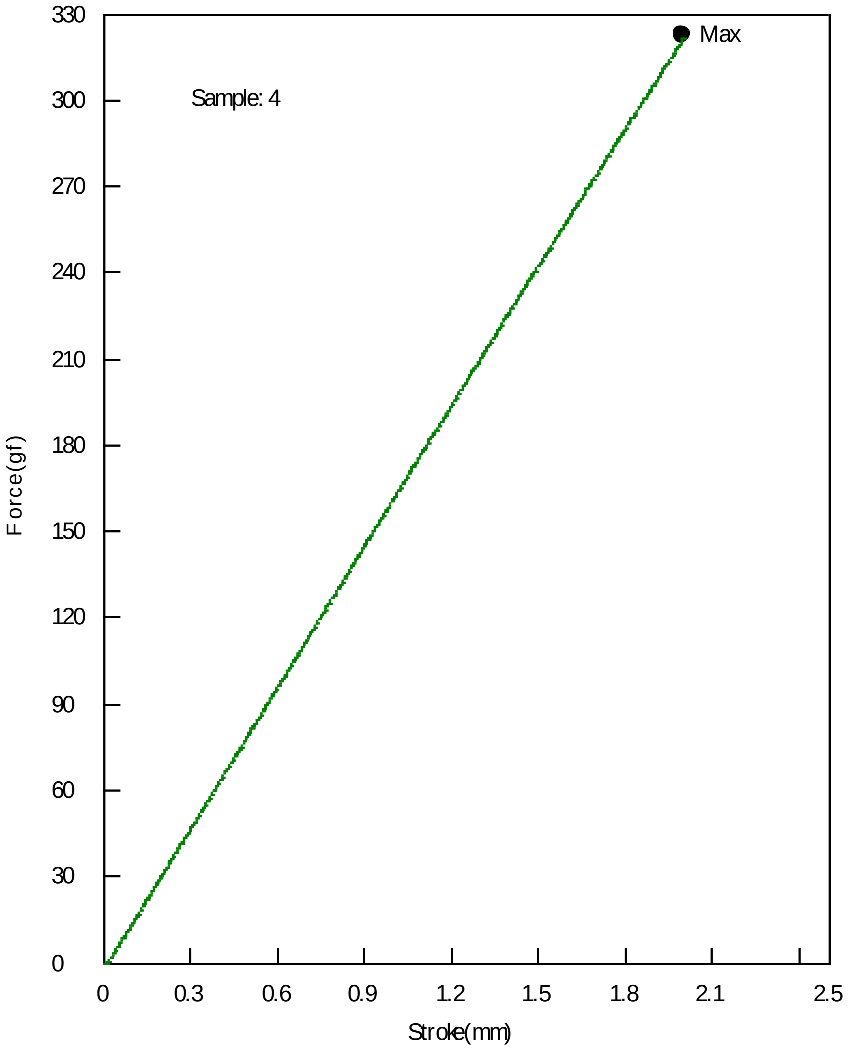
**Table9.** Paired '**t**' test for Tear drop loop between computer stimulation and experiment.

| Activation ( mm) | Experimental value for Tear drop loops(N) | Computer simulation value (N) |
|---|---|---|
| 0.5 | 0.827 | 0.809 |
| 1.0 | 1.584 | 1.618 |
| 1.5 | 2.261 | 2.426 |
| 2.0 | 3.298 | 3.236 |

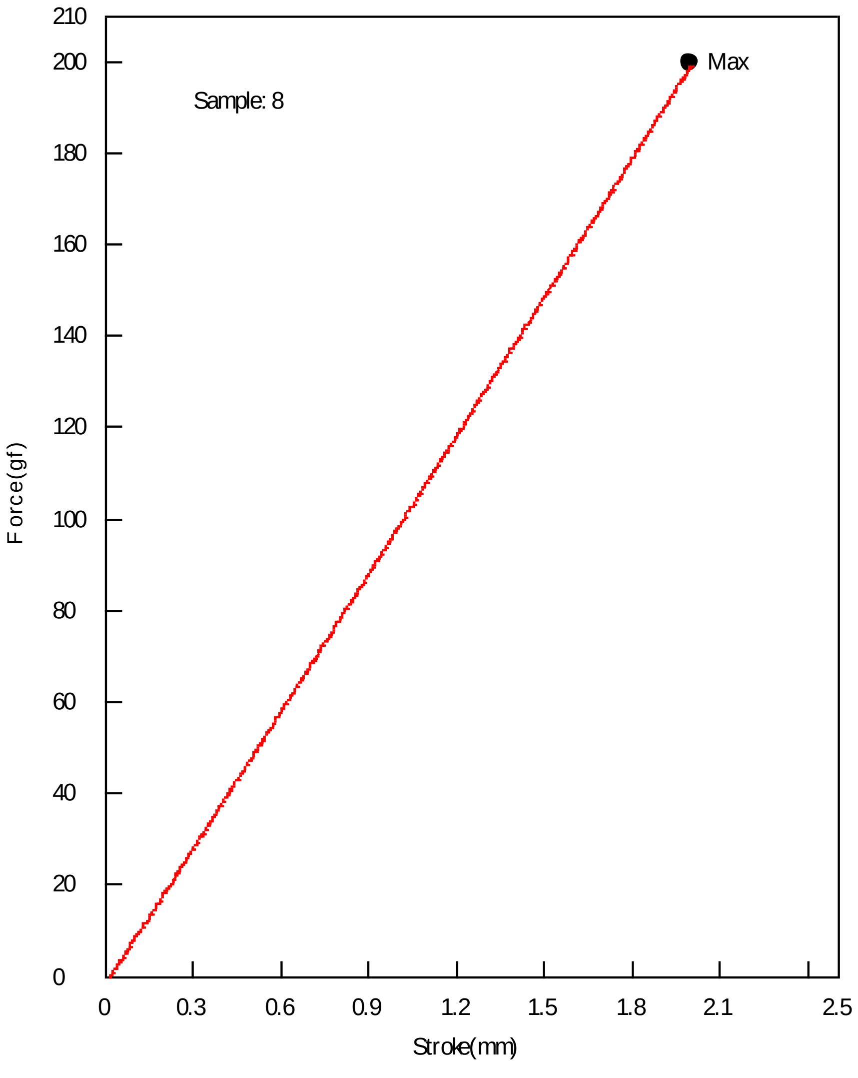
*P value = 0.5873 (not statistically significant.)*

# GRAPHS

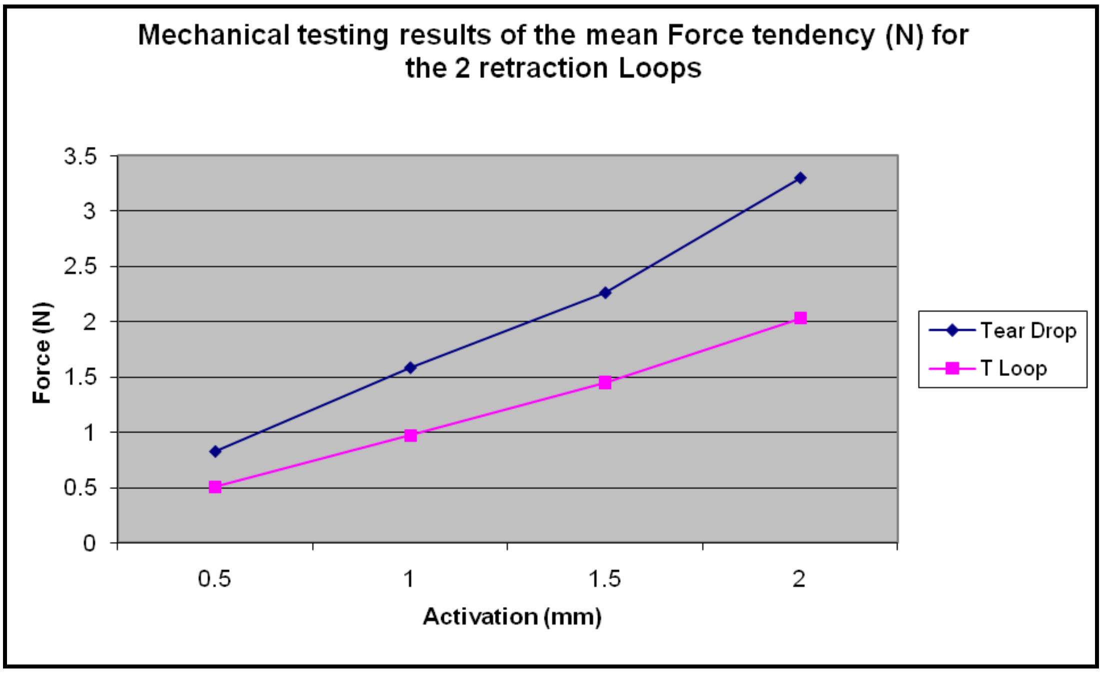
## Graph- 1: Force obtained on activation of Tear drop loop to 2mm in universal testing machine



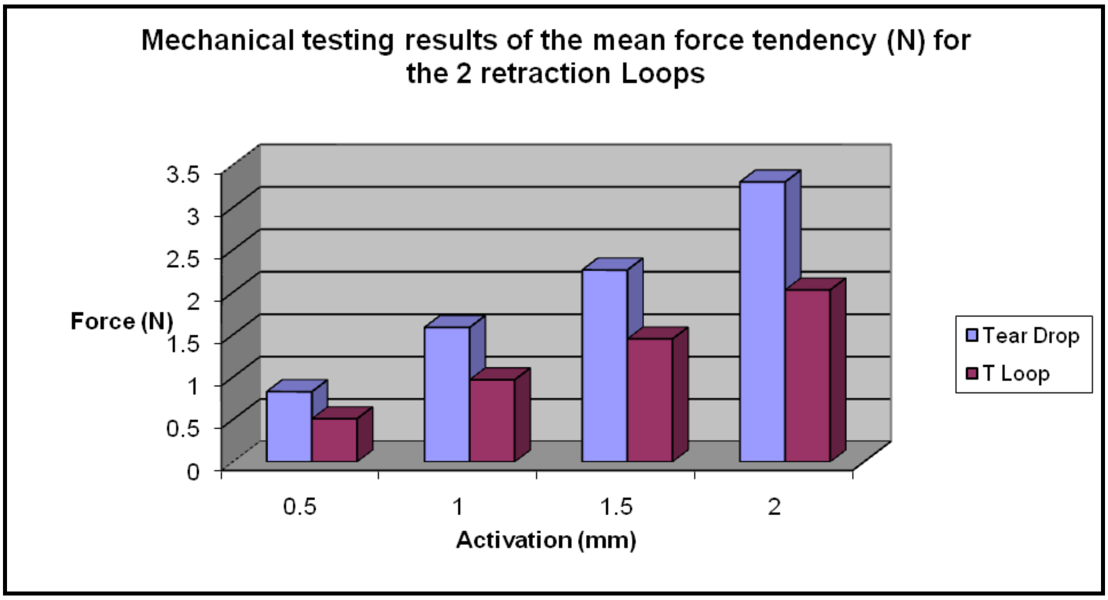
**Graph – 2: Force obtained on activation of T loop to 2mm in universal testing machine**



### Graph- 3: Mean force tendency (n) for T loops and tear drop loops (mechanical testing)




### Graph-4: Mean force tendency (n) for T loops and tear drop loops (mechanical testing)



# DISCUSSION

Biomechanics is the study and analysis of mechanical function in living bodies and the effect of the force on the form and motion of living bodies. Forces are load or external influences applied to a body that changes or tends to change the position of that body. If a force is applied to the center of the resistance the whole body moves equally in the direction of force applied. When orthodontic forces are applied and when the line of the force does not pass through the center of the resistance a moment is created.

In orthodontic mechanotherapy, several orthodontic mechanisms were designed for closing loops. Among the various devices described there are many ranges of loops, which may be used, incorporated into continuous or segmented arches, for vertical and saggital dental movement. Due to the large number of options, a great deal of attention must be paid when selecting the most appropriate model for each case.

In this choice, certain variables must be analyzed, among them the loop design, it quantity of activation, wire thickness, the metal alloy used, type of movement desired and the quantity of force necessary. When using loops for closing spaces, it is of the utmost importance for the professional to determine precisely the system of forces generated; that is, it is important for the orthodontist to know the magnitude of Orthodontic forces and the moments released when these devices are activated.

Orthodontic forces applied by means of orthodontic appliances are dependent on various factors that determine the success of the treatment. The operator has control over a few factors like moment: force ratio, magnitude of force and constancy of force **Burstone (1966)**[10]

Optimum control of tooth movement requires application of specific orthodontic force systems. Therefore, the knowledge of the mechanics of orthodontic appliances is essential to achieve desirable and predictable treatment results. **(Burstone 1966)**[10]

Load deflection rate is also important for an orthodontic spring as its moment to force ratio. Force produced per unit deflection is the load deflection rate. A low load deflection rate is preferred for orthodontic springs **(Poowadon Koson –ittikul et al. 2008)**[44] Low load deflection is preferred for two reasons: 1) it maintains a desirable force level in the PDL 2) It offers greater accuracy in controlling force magnitude. Hence an ideal orthodontic spring should have a greater M/F ratio and a lower load deflection rate**. (Burstone 1966)**[10]

Crown tipping, translation and root movement are examples of different types of tooth movement that can be produced with proper moment to force ratio **(Graber, Vanarsdall)**[22] An important aspect of orthodontic treatment is to understand tooth movement in response to mechanical loads and the associated adjacent tissue response at both clinical and histological levels **(Proffit)**[65]**.**

To move teeth in a controlled fashion, correct mechanical principles and an ideal orthodontic appliance must include adaptation of the device to the various types of malocclusion to better align the teeth to be moved and ease of placement in the mouth. While the final goal of a retraction system is efficient, effective closure of space within the dental arch, the designs and materials of the appliances used to accomplish this effect vary considerably.(**M.G.Faulkner et al 1991)**[33]

To provide the appropriate force system, the appliance must have the following mechanical characteristics: 1.It must provide appropriate levels of force and moment-to-force (M/F) ratios to achieve the tooth displacement desired.2. It must be able to undergo a reasonable range of activation/deactivation in which the appliance delivers relatively constant forces and moments.3. It must be small enough to fit comfortably in the space available for intraoral treatment. **(M.G.Faulkner et al 1991)**[33]

Number of procedures is used for retraction of anterior segments in treatment of extraction case. In sliding mechanics, force dissipates due to friction which is unknown and unpredictable. To counteract these undesirable effects the retraction of anterior teeth by frictionless system is based upon the incorporation of loop. **(Charles Burstone et al 1976)**[5]

In frictionless mechanics, teeth are moved without the brackets sliding along the arch wire. Retraction is accomplished with loops or springs, which offer more controlled tooth movement than sliding mechanics. The force of a retraction spring is applied by pulling the distal end through the molar tube and cinching it back. The moment is determined by the wire configuration and by the presence of pre activation or gable bends, which produce an activation moment.

The addition of helices lowers the load/deflection rate without significantly affecting the moment to force ratio (**M.G.Faulkner1991**)[33]. Reducing the load deflection rates of orthodontic springs is important for it provides a relative constancy of the moment- to- force ratio applied to the teeth with concomitant , force a stable dental ,movement (Stanley Braun et al 1997). A vertical loop's moment to force ratio can be improved by increasing its height, horizontal length, and loop diameter of the spring. The same is true of a T-loop, but once the horizontal section becomes equal in length to the vertical section, no improvement in the moment to force ratio is gained by lengthening the horizontal section**. (Burstone et al 1976)**[5]

Biomechanical consideration requires that archwire stiffness be an important criterion upon which rests the relationship between orthodontic forces and deflection within the elastic working range. Stiffness is directly related to the cross sectional size and shape of the wire and affected by bracket width, interbracket distance, the length of the wire, wire material, hardness and the incorporation of loops**.**

**Burstone et al 1982**[6] concluded that the most important consideration in the clinical use of attraction springs are the amount of distal activation, the angulations differential between the anterior and posterior teeth and centricity or eccentricity of the loop.

Loop design lead to a more efficient, hygienic, and comfortable mechanism for space closure. Principles of loop design include use of a rectangular wire for preventing the rolling of loop arch wire in the bracket slots, use of a simple design, a failsafe mechanism and adequate range of action to deliver continuous controlled force**. (Wick Alexander)**[64]

The different biomaterials used in orthodontic wires can be stainless steel, nickel titanium or beta-titanium. When formed into different configuration, they generate biomechanical forces which are transmitted to the appliance to produced tooth movement. The ideal arch wire should deliver a low constant force, good spring back and biocompatible**. (Nanda)**[46] β - Titanium wires have improved values of spring back which markedly increases their working range for tooth movement. For a given cross section, it can be deflected approximately twice as far as stainless steel wire without permanent deformation (**M.G.Faulkner)**[34]. The high formability of β-titanium allows the fabrication of closing loops with or without helices. (**Burstone AJO 1980)**[9]

Deformation results from stresses within the continuum induced by external forces or due to changes in its temperature. The relation between stresses and induced strains is expressed by **Hooke's Law (kannth)**[29]. The internal stress and strain in the orthodontic wire is responsible for the force delivery of the orthodontic wire. The stress strain ratio is one of the major factor in determining the resiliency and spring back of the given arch wire. Stress relaxation plays a major role in force decay. The amount of stress and strain in a cold worked area of the wire and the stress relaxation, possibly along with Bauschinger like effect determines the force delivery given by the activated orthodontic wire.

Design alterations in orthodontic appliances and the addition of helices or changes in alloy composition and processing are commonly used to allow clinicians to more accurately achieve desired forces for tooth movement in various clinical scenarios.

Therefore, accurate prediction of mechanical behavior as a function of shape and material properties is necessary in clinical practice. Finite element modeling is a powerful analytical technique for calculating stresses and strains within mechanically loaded structures. The method can be used to model intricate structures consisting of various shapes and materials under complex loading.

The finite element method is a numerical analysis technique used by engineers, scientists, and mathematicians to obtain solutions to the differential equations that describe, or approximately describe a wide variety of physical (and non-physical) problems[43].

Finite element analysis (FEA) has been mainly applied in orthopedic research for the evaluation of mechanical responses of bony structures to applied external forces. (**Retian)**[51] This method is particularly useful when several forces are applied to objects of complex shape and varied material properties. This method is based on the separation of the analysis shape into subdomains through finite elements. This separation allows a point analysis of the physical behavior of the object under varied loading conditions **(marina elisa Coimbra 2009)[38].**

The present study evaluated computer simulation to predict the force obtained after activation of Tear drop and "T" loops by comparing with mechanical testing method. Since principles of spring design are mostly satisfied by T loops and Tear drop loops, they were tested in this study.

The study was conducted using 0.0175 x 0.025 TMA loops. Two different loops were analysed in finite element method using ANSYS software (version 11) The structural analysis was carried out by subdividing entire loop structure into a smaller finite structure called finite elements. The boundary conditions were defined as similar to universal testing machine and study made to find load required to make deflection[38].

Fifty closing loops made of TMA 0.017 × 0.025-inch rectangular wires were bent. Based on the design of the loops, they were divided into 25 Tear drop loops and 25 T loops. Dimension of Tear drop loop was 7 mm height and 2.5mm internal arch circumference[38]. T loop was 5mm in anterior (α) vertical height and 4mm in posterior(β). Posterior Vertical height was made to be in level with auxilliary tube[37]. Width (horizontal length) of loop was 10mm and circumference is 2 mm. To avoid inter operator variability, the same operator prepared the specimens using a template with a light wire plier with TC tip (Dentronix USA) and tweed loop forming plier (Rocky Mountain, GERMANY). After preparation, each loop had 28 mm of total length of which anterior and posterior 5mm of wire allowed its attachment to a universal testing machine. Remaining 18mm represented the standard inter bracket distance (IBD) and loops are placed centrally. Pre activation bends were not given in any loop.

**Evaluation with Universal testing machine:**

The springs were then subjected to a tensile load on the universal testing machine. (SHIMADZU model: AG-IS 50 KN , JAPAN). A load cell of 20.0 N was used **(marina elisa Coimbra 2009)**[38]. For this purpose, one end of the specimen was fixed to the machine, and the other end was displaced. The loops were subjected to activation steps with increments of 0.5 mm at a rate of 1.0 mm per minute up to a maximum displacement of 2mm.The forces were measured.

**Evaluation using Computer simulation:**

The geometry of the springs used in the mechanical testing was obtained with the help of a jig. The different loop models were created in ANSYS. According to the characteristics of the loop structures and also considering the specific movements imposed by the intended mechanical activation, BEAM 4 elements were used for constructing loop models for analysis. This (finite) uni axial element can respond to tension, compression, traction, and torsion movements. The BEAM 4 element has 6 degree of freedom, and 3 translation and 3 rotation points around the axis. The modulus of elasticity of the orthodontic wire used in the FEA analysis of the springs was 71.70GPa with a Poisson coefficient of 0.130. All loops were discreted into finite elements, and an average of 237 elements was used for modelling. To simulate the activation similar to universal testing machine, 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. Force was determined for each loop during 0.5mm incremental activation up to 2mm of activation. Force value at various level of activation (from 0.5mm to 2mm) of T loop and Tear drop loop for the 25 samples were mechanically obtained and compared with the Computer simulated values.

Un paired 't' test was used for comparing 'T' loop and Tear drop loop results Paired 't' test was used for the comparison between the computer simulated and mechanical groups.

The results indicated that the Tear drop loop had the greatest force value at all levels of activation than the 'T' loop. These results are comparable with those of Burstone and Koenig who observed the relationship between the addition of helicoids and decreased force for various activation. Since the 'T' loop had greater wire incorporation than tear drop loop, It delivered less force for all activations. When possible, springs that release low force levels are preferred.

The optimal force for canine retraction has been suggested as 150gms to 200gms or 1.53N to 2.04N. **Marcotte**[37] has given that the retraction force for canine should not be more than 300gms. The results of the present study show that the 'T' loop at 1.5mm of activation and Tear drop loop at 1mm of activation will give the optimum force for canine retraction.

The computer simulation also presents similar results, indicating that the 'T' loop at 1.5mm of activation and Tear drop loop at 1mm of activation will provide the optimum force for canine retraction.

Student 't' test comparison between 'T' loop and Tear drop loop shows that at 0.5mm of activation 'T' loop gives a force of 0.51N of force and Tear drop loop gives a force of 0.826N.At 1mm of activation 'T' loop gives a force of 0.96N and Tear drop loop gives a force of 1.55N.At 1.5mm of activation 'T' loop gives a force of 1.46N and Tear drop loop gives a force of 2.24N.At 2mm of activation 'T' loop gives a force of 2.05N and Tear drop loop gives a force of 3.3N.The P value for the corresponding activations are less than 0.0001 indicating extremely significant difference between the 'T' loop and Tear drop loop. This indicates that the 'T' loop will deliver less force and greater M/F ratio at all activations compared to the Tear drop loop.

Paired 't' test comparison between the mechanical and computer simulated group for the 'T' loop gives a P value of 0.1915 indicating that the difference between the group is statistically insignificant. The findings of this study go in accordance with the study of **Maria Elisa**[38]

Paired't' test comparison between the mechanical and computer simulated group for the Tear drop loop gives a P value of 0.5873 indicating that the difference between the groups is statistically insignificant. The findings of this study goes in accordance with the study of **Maria Elisa**[38]

The present study indicates that the force value measured by mechanical means and by computer simulation does not have any significant difference between them. The study also indicates that the 'T' loop gives less force values for all activations compared to the Tear drop loop[5]. A high correlation coefficient of T loop (0.9952) and Tear drop loop (0.998) was found between mechanical testing and Computer simulation (finite element analysis).

## FUTURE PROSPECTS

There are several future studies stimulated by current investigation. Only two types of loops were tested in the present study. Considering the diverse number of retraction loops that are clinically used in orthodontics, software simulated studies can be performed to evaluate more mechanical properties like torsion, moment, moment: force ratio, load deflection rate. This study may lead to development of customized software for orthodontic application and choice of appliance use in clinical practice.

# SUMMARY AND CONCLUSION

The present study was conducted to evaluate the force obtained after activation of Tear drop and “T” loops by mechanical method and by finite element method. 25 ‘T’ loops and 25 Tear drop loops were made and activated incrementally. Computer simulation (Finite element analysis) of the Tear drop loop and T loop were done using ANSYS software and the force exerted was evaluated. The values obtained in both the methods were compared and analyzed. The results show that the ‘T’ loop delivered statistically less force for all activation compared to Tear drop loop. Stress concentration for the Tear drop loop was maximum in the curvature of loop[38].(fig 18)

The results also showed that there was no statistically significant difference between the computer simulation and mechanical measurement signifying the accurate prediction of Finite element analysis. Since computed simulated technique accurately predicts the experimentally determined mechanical behavior[38] of T loop and Tear drop loop, it shall be consider as alternative for designing orthodontics appliance before treatment.

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