EVALUATION OF MOMENT-TO-FORCE CHARACTERISTICS OF PREACTIVATED SYMMETRICAL T-LOOP MADE OF TWO DIFFERENT ALLOYS: AN IN-VITRO STUDY

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CERTIFICATE

This is to certify that this dissertation titled " **EVALUATION OF MOMENT-TO-FORCE CHARACTERISTICS OF PREACTIVATED SYMMETRICAL T-LOOP MADE OF TWO DIFFERENT ALLOYS: AN IN-VITRO STUDY.**" is a bonafide record of work done by **Dr.SABITHA R.NAIR** under my guidance during her postgraduation study period between 2009-2012.

This dissertation is submitted to **THE TAMILNADU DR. M.G.R. MEDICAL UNIVERSITY,** in partial fulfillment for the degree of **Master of Dental Surgery** in Branch V-Orthodontics and Dentofacial Orthopaedics.

It has not been submitted (partially or fully) for the award of any other degree or diploma.

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CONTENTS

INTRODUCTION

The cascade of biological events that induce orthodontic tooth movement is initiated by mechanical stresses in the periodontium. Forces are transferred to the teeth by the clinician using appliances designed to displace teeth a prescribed amount in a desired direction. This sends signals to the cells to remodel tissues in a way that allows teeth to move. To interpret the biological responses to activation of any orthodontic appliance, each interface in the process must be thoroughly understood.

Smith and Storey [64] stated that maximum amount of tooth movement would occur at some optimal force level. They theorized that rate of tooth movement increases with force up to a point, after which the rate decreases or ceases as force levels continue to increase. Proponents would argue that above the optimal force, greater forces prevent the recruitment or differentiation of cells or that the high pressures cause tissue hyalinization, slowing tooth movement and affecting cell-tissue interactions.

However, one of the inherent limitations of orthodontics is the point of force application on the bracket, which is always at a distance from the center of resistance of the tooth or a consolidated section of teeth. To overcome this limitation and achieve the desired tooth movement, a counter-couple or moment is required.

The relationship between the applied force system and the type of movement can be described by the moment–to-force ratio. M:F ratio at the center of resistance determines the stress pattern in the periodontal ligament and how a tooth or a segment of rigidly connected teeth will move.[44]

Space closure represents a fundamental stage in considerable number of cases. It can be done either by friction or frictionless mechanics. In friction method, teeth slides by force application and will be guided by continuous archwire. The possible side effects of friction mechanics are tipping, binding of archwire, lack of vertical control, risk of anchor loss and incisor extrusion. The drawbacks of sliding technique can be overcome with a frictionless system including a loop as a source of force.

Burstone [13] developed T-Loop for canine retraction with TMA wires. The major advantage was its low load-deflection rate and large spring back. This enabled an optimal magnitude of force for canine retraction. However, it was not ideal to produce enough moment to force ratio for canine translation.

The load-deflection rate is dependent on three fundamental factors. (1)Wire cross section, (2)Wire material and (3)Wire length.[8] The advances in orthodontic wire alloys have made it possible to control wire stiffness by varying material properties-namely, the modulus of elasticity, which is known as “variable modulus orthodontics’’.

Austenitic nickel-titanium (NiTi) wires are low modulus alloys. The principal advantage of nickel titanium archwires is that they can be deformed over long distances, while continuing to deliver clinically acceptable force levels. These wires provide a relatively low constant force over a long range of action. Superelastic nickel titanium wires can be significantly loaded without permanent deformation and exhibit shape memory.[57] These wires also show a characteristic unloading plateau (hysteresis). Therefore it appears to meet material requirements for closing loops, aiming to provide a constant and ideal M:F ratio.

However, studies have shown that non preactivated NiTi closing loops failed to achieve optimum M:F ratio for tooth translation.[54] The appropriate preactivation gable bends might provide a sufficiently high M:F to retract a single tooth or a group of teeth via translation.

The low formability of nickel-titanium alloy limits its applications where considerable bending of an appliance is required. By heating a NiTi archwire in the martensitic phase to 300-520°c, a new shape can be programmed into it.[58] In 1990, **Sander** introduced a pulsed heat–induction method for reprogramming the memory of superelastic nickel titanium wires for specific clinical purposes. The Memory-Maker permanently changes the conformation of superelastic nickel titanium wires without destroying their superelastic properties.[57]

Little has been written on the M:F generated by closing loops made of superelastic NiTi wires. Therefore the purpose of this study was to evaluate the efficiency of T-loops with different preactivations constructed from superelastic NiTi wires and to compare it with those of TMA wires.

AIMS AND OBJECTIVES

The aim of this in-vitro study was to investigate the Moment-to-Force ratio (M:F) generated during the deactivation of closing loops(T-LOOP) constructed from TMA and NiTi wires in-corporated with preactivation bends of 0°,15° and 30°.

This study compares the M:F generated by

1. Closing loops fabricated from two different materials- NiTi and TMA of same design and angulations.
2. Closing loops fabricated with different degrees of preactivation : 0º, 15º and 30º.
3. Closing loops fabricated from NiTi wires using two different methods- Furnace heating and Pulsed heat induction method.

Hypothesis tested:

1. By increasing the pre-activation bends, the M:F increases for all the closing loop specimens during the deactivation phase.
2. The closing loop specimens made of TMA and NiTi with similar pre-activation bends produces statistically significant difference in the M:F.
3. NiTi closing loops fabricated using two different methods with similar pre-activation bends, produces no statistically significant difference in the M:F generated.

REVIEW OF LITERATURE

Schwarz [61] (1932) proposed the classic concept of the optimal force. He defined optimal continuous force as “the force leading to a change in tissue pressure that approximated the capillary vessels’ blood pressure, thus preventing their occlusion in the compressed periodontal ligament.’’ According to Schwarz, forces well below the optimal level cause no reaction in the periodontal ligament. Forces exceeding the optimal level would lead to areas of tissue necrosis, preventing frontal bone resorption. Tooth movement would thus be delayed until undermining resorption had eliminated the necrotic tissue obstacle.

Oppenheim[47] (1942) advocated the use of the lightest force capable of bringing about tooth movement. The optimal force for tooth movement may differ for each tooth and for each individual patient.

Smith and Storey [64] (1952) stated that maximum amount of tooth movement would occur at some optimal force level. They theorized that rate of tooth movement increases with force up to a point, after which the rate decreases or ceases as force levels continue to increase. Proponents would argue that above the optimal force, greater forces prevent the recruitment or differentiation of cells or that the high pressures cause tissue hyalinization, slowing tooth movement and affecting cell-tissue interactions.

Reitan [51] (1967) demonstrated cell-free compressed areas within the pressure site even in cases where light forces were applied and also advocated the use of very light forces.

Burstone and Koenig [15] (1976) stated that although it may be possible to design loops or retraction springs to deliver an adequate moment-to-force ratio for controlled tipping around the apex of an incisor or a canine, translator movements are not possible, considering the intraoral limitations on spring height. This can be overcome by the placement of gable bends or angulation in a loop or retraction spring. The M:F resulting from activating various loop configurations are insufficient to prevent uncontrolled tipping unless Gable bends are included.

If a higher moment-to-force ratio is required, it is necessary to use a closing loop with greater vertical height. Additional horizontal wire placed gingivally will also raise the moment-to-force ratio for any given loop length. At the same time, load-deflection rate is also lowered.

Burstone and Pryputniewicz [16] (1980) designed an in vitro study using laser holography to establish the required force system applied on the crown of a maxillary incisor that would produce different centers of rotation, as in lingual tipping, translation and root movement. It was found that the center of resistance was at a point one-third of the distance from the alveolar crest to the apex.

Burstone and Goldberg [13] **(1980)** presented the clinical applications of beta titanium. It is a material that has an excellent balance of properties, including high springback, low stiffness, high formability, highly ductile, and can be welded together without appreciable reduction in the mechanical properties. The springback for Beta-titanium is superior to that of stainless steel and can be deflected almost twice as much as stainless steel wire without permanent deformation. It delivers force values less than half that of stainless steel. Beta titanium has a modulus of elasticity that is less than that of stainless steel and about twice that of nitinol. This makes its use ideal in situations in which forces less than those of stainless steel are necessary and in instances in which a lower modulus material such as nitinol is inadequate to produce the desired force magnitudes.

Burstone [11] **(1981)** presented a new approach, Variable-Modulus orthodontics which bases force-magnitude control on varying primarily the material rather than the cross section of the wire. The advantages of variable-modulus orthodontics includes better control over the amount of play between attachment and wire, orientation of wires for directional distribution of forces, preferential orientation of rectangular wires, and over-all reduction in the number of wires used for treatment. These wires can be simple in design, so-called straight wires, or more complicated in configuration, incorporating loops.

Hoecevar [28] (1981) reported that when a tooth is subjected to a tipping moment, strain is concentrated in the areas of the alveolar crest and root apex. Thus a light force can tip a tooth readily, while translation, involving a more even distribution of strain throughout the length of the root, requires more force with little or no moment.

Burstone [12] (1982) described the clinical application of frictionless attraction springs using the segmented arch technique. The material used was beta-titanium. The wire cross sections were kept as small as possible, limited by the moments needed rather than the force. Additional wire should be placed as far apically as possible to increase the activation moment-to-force ratio. The loop centricity affects the rate of change of the moment-to-force ratio in the alpha and beta positions. The 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. In addition, it adds to the accuracy of determining the force system, since small errors in the shape or geometry of the spring will not change the forces produced.

Kusy and Greenberg [36] (1982) compared beta titanium and nickel titanium wires, found that stiffness of the two alloy compositions overlap substantially except for those wires with the lowest and highest stiffness that is the 0.016 and 0.018 NiTi and the 0.017 x0.025 and 0.019 x0.025 beta titanium archwires respectively.

Schwaninger [56] (1982) examined the effect of corrosion of flexural properties of Nitinol wires immersed in 1% sodium chloride for 11 months. Corrosion does not effect the physical properties of the wire and the early failure of control and test wire is due to presence of surface defects during manufacturing and not due to the effect of corrosion.

Drake et al [22] (1982) conducted a study on the mechanical properties of three sizes of stainless steel, Nickel Titanium and Titanium molybdenum orthodontic wires in tension ,bending and torsion. In tension stainless steel had the least maximum elastic strain or spring back, whereas titanium molybdenum had the most. In bending and torsion, the SS wires had the least recoverable stored energy, whereas the Nickel titanium wires had the most. Titanium molybdenum teardrop closing loop delivered less than one half the force of a comparable stainless steel loop for similar activation.

Smith and Burstone [58] (1984) explained basic relationships between forces and tooth movement and their potential for clinical relevance. Forces produce translation, 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. The tendency to rotate is due to the moment of the force, which is equal to force magnitude multiplied by the perpendicular distance of the line of action to center of resistance. 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. A moment toforce ratio of about 8:1 produce controlled tipping, 10:1 result in translation of tooth and 12:1 produce root movement.

Burstone et al [17] (1985) conducted a study on NiTi by means of a bending test to determine wire stiffness, spring back and bending moment. He observed that Chinese NiTi wire has an unusual deactivation curve in which relatively constant forces are produced over a long range of action. At large activations, NiTi wires have a stiffness of only 7% of stainless steel wire and at small activations 28% of stainless steel wire. For same activation, the forces produced were 36% of Nitinol wire. Chinese NiTi also demonstrated phenomenal spring back. It can be deflected 1.6 times as far as stainless steel wire without appreciable permanent deformation.

Gjessing [26] (1985) described a canine retraction spring constructed from 0.016x 0.022 inch stainless steel. Alpha moments are moments generated by the mesial part of the spring and acting at the canine, whereas beta moments act on the anchorage teeth. The beta moment is a desirable complement to the alpha moment, as it reduces mesial movement of the anchorage tooth group by inducing translation. Gjessing advocates incorporation of a segment of a circle in the distal leg of the spring to eliminate undesirable beta moments acting at the second premolar bracket and tending to move the root apex too far mesially.

Optimum force provides the periodontal tension which generates maximum cellular and biochemical activities responsible for tooth movement.

Extension of the load beyond this level can lead to root resorption, loss of anchorage, and alteration of the M:F. It was verified that active appliance components can be precalibrated with tested and documented characteristics for controlled tooth movement.

Kusy and Tulloch [37] (1986) analyzed orthodontic tooth movement by means of the center of rotation model and the concept of moment to force ratios. According to them several equivalent force systems are considered at both the bracket and the center of resistance of the tooth. When moment to force ratios are evaluated at the bracket, the laws of physics appear to be suspended: inconsistencies occur as single forces applied at different points claim equivalent results and pure translational movements purport to be nonzero moment to force ratios.

Miura et al [43] (1986) conducted a study on Japanese NiTi alloy and observed that the alloy wire delivered constant forces over an extended portion of the deactivation range. Japanese NiTi alloy wire was the least likely to undergo permanent deformation during activation. Heat treatment enabled the load magnitude in the superelastic region to be influenced and controlled by temperature and time. Hence an archwire delivering various magnitudes of forces over a given activation could be fabricated from the wire of the same diameter. The superelastic property of nickel titanium wires has been attributed to a phase transformation from the body-centered cubic austenitic form to the hexagonal close-packed martensitic form of NiTi when the stress reaches a certain level during activation. Upon deactivation, the reverse–phase transformation from the martensitic to the austenitic structure takes place when the stress is decreased to an appropriate level.

Vanden Blucke [62] (1987) studied the location of the centers of resistance for various symmetric units of the anterior maxillary dentition for a lingually directed force in two dry human skulls. The instantaneous center of resistance for the rigidly fixed six anterior teeth was located at ±7mm apical to the interproximal bone level. With a unit of six anterior teeth, the apical shift of the center of resistance was the greatest. Increasing force levels had little effect on the location of the center of resistance of a given unit. This phenomenon was observed in both the skulls tested, suggesting that general trends may exist in the displacement characteristics of the dentition when subject to controlled force systems.

Buckthal et al [10] (1988) studied the effects of disinfectants like 2% glutaraldehyde, chlorine dioxide and iodophor on mechanical properties and surface topographies of 0.017x0.025 inch Nitinol and Titanal wires. No significant changes were detected in the fundamental stiffness or inherent strength of the wires after multiple disinfectant cycles.

Harris et al [27] (1988) conducted a study on mechanical properties of Nitinol in simulated oral environment, at various levels of acidity and at different levels of static deflection. He noted that there is significant decrease in mechanical properties of incubated wires compared with control group kept dry and unstressed. Long-term use of a nitinol wire would appear to be associated with decreased performance, particularly in the elasticity of the wire for which it is noted orthodontically.

Tanne, Koenig and Burstone [60] (1988) developed a three dimensional FEM model for the upper right central incisor on the basis of average anatomic dimensions. The center of resistance and centers of rotation were determined for varying M:F applied at the midpoint of the crown. The center of resistance was located at 0.24 times the root length measured apical to the level of alveolar crest.

Burstone et al [14] (1989) have shown that the neutral position of the loop configuration is altered by the introduction of Gable bends. The neutral position can be defined as the horizontal separation of the vertical legs of the spring before the introduction of a horizontal or mesiodistal force.

Ronay et al [49] (1989) reported that the interbracket position of the bend is crucial for the force system delivered to the two teeth, be they two adjacent teeth or, as in the segmented technique, units of teeth with a larger interbracket distance. If the interbracket distance is divided into thirds and if the position of the V bend is in the central third, it will provide two equal and opposite moment. With increasing eccentricity, the moment on the tooth closest to the bend will be bigger and smaller at the distant tooth. If the V bend is not centered but still within the middle third, two opposite moments of different magnitudes will be generated and vertical forces will arise to establish equilibrium. When the eccentricity of the bend reaches the situation where the V bend is exactly one third of the interbracket distance from one of the teeth, the moment on the distant tooth is reduced to zero and only intrusive force will exist. This point has been referred to as the point of dissociation.

If the V bend is moved further eccentrically so that its distance to one of the brackets is less than one third of the total interbracket distance, two moments of different magnitude but of the same direction are generated. With increasing eccentricity, the moments will approximate the same size and the vertical forces generated will increase.

Kapila et al [31] (1989) described the mechanical properties and clinical applications of several wires used in orthodontics. The advances in orthodontic wire alloys have made it possible to control wire stiffness by varying material properties-namely, the modulus of elasticity, which is known as “variable modulus orthodontics’’. Nitinol wires have good spring back and low stiffness. Higher springback values provide the ability to apply large activations with a resultant increase in working time of the appliance. This, in turn, implies that fewer arch wire changes or adjustments will be required.

Nitinol wires have a larger recoverable energy than stainless steel or beta titanium wires when activated to the same amount of bending or torquing. Beta Titanium has a modulus of elasticity that is less than that of stainless steel and about twice that of Nitinol.

Beta Titanium wires deliver about half the amount of force as do comparable stainless steel wires. Beta Titanium wire provide adequate springback, average stiffness, good formability and joinability. Although wire characteristics determined by these tests do not necessarily reflect the behavior of the wires under clinical conditions, they provide a basis for comparison of these wires.

Michael R Marcotte [42] (1990) stated that assuming a constant force level, the best way to control the M:F is one of the following ways:

1. For translation, adjust the moment at the bracket, remembering that the necessary moment value is F x D (D = distance from bracket to Center of resistance.)
2. For controlled tipping, reduce a little of the moment.
3. For root movement, add more moment to the bracket.

It is better to change the M:F by increasing or decreasing the moment values at the bracket than by changing the force value.

Kapila [32] (1991) evaluated the effect of clinical recycling on mechanical properties of NiTi alloy wires and observed that recycling produces significant changes in both the loading and unloading characteristics of NiTi wires. Scanning electron microscopy demonstrated pitting on both Nitinol and NiTi wires.

Khier et al [34] (1991) showed that heat treatment of superelastic wires at 500°c for 10 minutes had minimal effect, whereas heat treatment at 500°c for 2 hours caused decrease in the average superelasticity. Heat treatment at 600°c resulted in loss of superelasticity. The differences in the bending properties and heat treatment responses of wires were due to the relative proportions of the austenitic and martensitic forms of nickel-titanium alloy (NiTi) in the microstructures of the wire alloys. During loading both superelastic and non-superelastic wires behave in a similar manner but during unloading superelastic wires exhibit horizontal slope at constant moment values. But nonsuperelastic wires had much greater slopes compared to superelastic wires. Another difference was superelastic wires showed an average of 10° to 15° permanent deformation compared to nonsuperelastic wires of 35° to 40°.

Faulkner et al [23] (1991) in his study clearly showed that the typical stainless steel vertical loop has two major limitations. First, its activation range is very restricted; second, the moment:force produced is also well below ideal if controlled tipping or translation is desired. While the use of alternate materials and cross sections can change the level of force and moment to a limited extend, the M:F remains unaltered. The changes in activation limits are also relatively small.

In comparison to the standard vertical loop, the preactivated loop produces larger activations without permanent deformation of the appliance.

As tooth movement occurs, the level of force remains at a more constant level than it would if the standard loop were used. The preactivation allows application of moments that are two to three times the original magnitude. The resulting moments are still not large enough to produce translation. If the proper gabling is coupled to the design with the helices, it produces higher M:F and are less sensitive to minor manufacturing and placement errors than the standard vertical loop. TMA is superior to stainless steel for the appliance design because of a large ratio of yield stress to the elastic modulus. For preactivations approximately above 12.5°, the spring would yield before reaching the neutral position.

Ronald H.Roth [48] (1991) explained the concept behind the use of double keyhole loop archwire mechanics which was introduced by John Parker of Alameda, California. 1) Double keyhole loop allow the operator the luxury of complete space closure with one set of archwires; 2) allow a reasonably happy medium between severe tipping and sliding mechanics; and 3) allow the operator to select how the space will be closed, from the front backward or from the back forward, and how much of which. In addition, the double keyhole loops control the canine rotation during extraction site closure and make handy elastic hooks for any arrangement of elastics that may become necessary, including midline shifting elastics.

Chen et al [19] (1992) conducted a study on Chinese NiTi wire and concluded that Chinese NiTi shows 100% recovery at 90° bending angle and has along constant range of bending and torsional moments. It also possesses low stiffness, high spring back and superelastic properties.

Smith et al [57] (1992) conducted a study on the effects of clinical use and various sterilization and disinfection protocols on three types of NiTi, Beta titanium and stainless steel wires. He concluded that clinically insignificant differences occur between new and used arch wires.

Barett et al [3] (1993) conducted a study to evaluate the corrosion rate of standard orthodontic appliances with Stainless steel and Nickel titanium arch wires and noted orthodontic appliances released measurable amounts of Nickel and chromium when placed in artificial salivary medium. The Nickel release reached a maximum after 1 week and thereafter diminished with time while the chromium release increased during the first two weeks and thereafter leveled off during the subsequent weeks. For both the arch wire types the release of Nickel was 37 times greater than that of chromium.

Sangkyu Han et al [55] (1993) studied the degradation of Nickel titanium spring properties in simulated oral environment and stated that Nickel titanium springs suffered no degradation while stainless steel spring became slightly compliant to stretching and polyurethane elastics lost a large portion of their force generating capacity.

Hoenigl et al [29] (1995) determined the force system of a prefabricated and preactivated T-loop used for reciprocal space closure by simultaneously measuring the horizontal and vertical forces, as well as the moments using a computer controlled measuring apparatus. At a loop activation of 7mm, the anterior and posterior segments first underwent controlled tipping, then translation, and finally, root uprighting as the moment-to-force ratio increased with deactivation. After the loop has been deactivated to 4mm, it should be exchanged to avoid root abutment.

Bishara et al [4] **(1995)** conducted a study to compare the thermodynamic properties of Nickel-titanium orthodontic arch wire and concluded that shape memory effects can be regarded as a combination of thermoelasticity and pseudoelasticity and the recovery rate seems to gradually increase as the temperature reaches the upper limit of TTR.

Braun and Marcotte [7] **(1995**) stated that, since a relatively constant moment-to force ratio is required to maintain the center of rotation of the active unit, the appliance delivering the force system must have a low load - deflection rate with a long range of activation. The load-deflection rate is dependent on three fundamental factors. (1)Wire cross section, (2)Wire material and (3)Wire length. Altering the material affects the spring rate in direct proportion to its modulus of elasticity.

Pilon et al [45] **(1996)** studied the relationship between the magnitude of a constant continuous orthodontic force and rate of bodily tooth movement in young adult male beagle dogs. The conclusion of the study was that other factors than magnitude of force are involved in determining the rate of subsequent tooth movement. Individual differences in bone density, bone metabolism, and turn over in the periodontal ligament may be responsible for the variation.

Kusy [37] (1997) gave the following criteria for an ideal archwire - esthetic, good range, tough, poor biohostability, good springback, low friction, weldable, springy, formable, biocompatible, resilient and strong. He stated that specific wire will do some things well and others poorly, but no wire will do it all.

Kuhlberg and Burstone [35] (1997) studied the effects of off-center positioning on the force system produced by symmetric T-loop springs. He concluded that a centered T-loop produces equal and opposite moments with negligible vertical forces. 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. A standard shaped T-loop can be used for differential anchorage requirements by altering the activation and mesial-distal position of the spring.

Bourauel et al [5] (1997) showed superelastic retraction T loop spring was made from Ormco and GAC Sentalloy of different batches and evaluated by using the orthodontic measurement and simulation system. OMSS measures the force system of T loop based on a mathematical model and the motions by means of the computer controlled positioning stages. Ormco delivers highest distalizing force compared to sentalloy. Within Ormco or

Sentalloy each batch of NiTi alloys displays different superelastic behaviour. The superelastic retraction springs deliver constant forces and moments over a broad range of activation, have very low distalizing forces, very low force/deflection rates; therefore the danger of exceeding the biological or physiological limits is minimized.

Siatkowski [59] (1997) presented a systematic approach to closing loop design for use in continuous arch wires by using Castigliano's theorem, then refined, using FEM stimulations, and then verified experimentally. The result of this process was the Opus loop, which is capable of delivering a target M/F within the range of 8.0-9.1mm inherently, without adding residual moments. The experimental results show that the loops must be bent accurately to achieve their design potential.

Raboud et al [47] (1997) investigated lateral and occlusal force systems for various appliance designs. At maximum activation the vertical loops produce higher forces than the T springs. None of the vertical loop designs approach an M/F ratio that is thought necessary to promote translation. Relatively high forces are produced by total activations of only approximately 1.0 mm. Prebending the ends a total of 20^0 out of plane does produce occlusal moments, which at maximum activation will partially counteract the tendency of the buccally applied force systems to cause rotation around the longitudinal axis of the tooth.

The analysis of the T spring designs shows that the use of more material, coupled with the different material properties of the TMA, results in force systems closer to those necessary for translation. The elastic activations possible are also 3 to 4 times larger than with the stainless steel vertical loops. The out-of-plane bends necessary to counteract the tendency for longitudinal axis rotation are larger and therefore the designs are far less sensitive to inaccuracies in the actual clinical manufacture of these appliances. A total out of plane bend of 40º produces an occlusal M:F that should limit longitudinal axis rotation and also produces a planar M:F that promotes tooth translation.

Oltjen et al [44] (1997) conducted a study to determine the stiffness characteristics of several solid and multistrand Nickel titanium and Stainless steel orthodontic wires. They concluded that the wire stiffness can be altered not only by changing the size but also by varying the number of strands and the alloy composition. An important finding was the dependence of stiffness on deflection for the most of the wires measured.

Airoldi et al [2] (1997) conducted a study to determine the oral environment temperature changes induced by cold/hot liquid intake as it has a direct effect on the force delivered from superelastic Nickel titanium wires. It was found that the temperature changes induced in the oral cavity by liquid intake differ from site to site in a nontrivial way. The lower dental arch is exposed, on the whole, to a larger temperature change, while the upper dental arch is exposed to a minor change. In the temperature sensitive sites, as a consequence of liquid intake, the temperature suddenly changes, reaches a maximum /minimum value and, then, recovers the oral temperature in a nearly exponential way. The temperature shots were felt as force shots or force relaxations during hot or cold drinking respectively.

Demetrios [21] (1998) demonstrated that Moment:Force decreases as the height of the loop is decreased. A 20mm high T-loop is needed to achieve a M/F ratio of 10:1.This is not very practical. Another method is to bend the wire so that the free end of the loop is not parallel to the bracket slot even before the loop is activated. This is called preactivating the loop.

Demetrios [20] (1998) compared 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. In this study, the forces and moments produced by a straight portion of an archwire were transferred from the brackets to the center of resistance. The findings of this study were: The force systems developed by an ideal arch cannot be used directly to estimate tooth movement. They should first be transferred to the center of resistance of the teeth and the force systems at the center of resistance may differ significantly from the force systems at the brackets.

Meling et al [41] (1998) conducted a study to determine the influence of short term temperature changes on the force exerted by superelastic Nickel titanium arch wires activated in orthodontic bending. The activated specimens were subjected to cold (10°c) or hot (60°c) water under constant deflection, simulating an inserted arch wire that is subjected to cold or hot drinks or food during a meal. It was found that the conventional nickel titanium wire was marginally affected by temperature changes while the superelastic wires were strongly affected by short term application of cold or hot water. He concluded that the short term exposures to hot liquid increased the bending force exerted for a given deflection transiently. The effect of short term exposures to cold liquid was not always transient, the bending force remained sub-base line for a number of thermosensitive wires tested for a prolonged time.

Ferreira [24] (1999) determined the resultant loads after successive activations and the spring rate of double delta closing loops of different wire materials and cross-sections through tension tests. The springs were fabricated with stainless steel, cobalt chromium and titanium-molybdenum. The stainless steel springs showed the highest total average load followed by the cobalt-chromium springs and finally by the titanium-molybdenum springs. The titanium-molybdenum 0.017 x 0.025 inch (Ormco) springs displayed the lowest average load for each tested 0.5 mm of activation and also the lowest spring rate. The spring rate is dependent on wire material, crosssection, and spring design.

Jie Chen et al [18] (2000) demonstrated that the moments and forces generated by a T-loop spring are functions of its geometry and gable angle combined with heat treatment. In general, increasing its vertical or horizontal dimension reduces the load-deflection rate and the moment-to-force ratio. Gable preactivation and stress relieving heat treatment has the opposite effect.

Rudolph et al [51] **(2001)** investigated the types of orthodontic forces that cause higher stress, specifically at the root apex of the maxillary central incisor. The finite element analysis showed that purely intrusive, extrusive and rotational forces had stresses concentrated at the apex of the root. The principal stress from a tipping force was located at the alveolar crest. For bodily movement, stress was distributed throughout the PDL; more concentrated at the alveolar crest.

Lindauer [39] **(2001)** stated that the goal of orthodontic treatment is to move teeth a prescribed distance in a predetermined direction to enhance the esthetic and functional aspects of occlusion and achieve a stable result. The force systems produced by orthodontic appliances, consisting of both forces and moments, displace teeth in a manner that is both predictable and controllable.

Brantley [6] **(2001)** reported that despite the inherent excellent formability of beta-Ti wire, this wire processing can be problematic because of the reactivity of titanium that can result in some batches of beta-Ti wire being susceptible to fracture during clinical manipulation.

Braun et al [8] **(2002)** illustrated a simple means of preserving the neutral position of closing loops. Appropriate magnitudes and occlusogingival locations of the gable bends are shown to be vital to maintain the neutral position of the closing loop. He concluded that Gable bends should be distributed occlusogingivally in all loop configurations to achieve forecastable M/F ratios at the active and reactive teeth.

Brezniak et al [9] (2002) reported that light forces, generally less than 200gm produce adequate biologic responses in the periodontium and heavier forces are associated with hyalinization of the periodontal ligament, undermining bone resorption and are implicated in root resorption.

James J. Baldwin [30] (2003) explained that a single force acting on the crown of a tooth will have an equivalent force plus a moment acting at the center of resistance of the tooth. Teeth do not move directly as the result of force application, but rather through the change of stress in various parts of the periodontal ligament. The stress in the periodontal ligament is best considered from the standpoint of forces related to the center of resistance.

Thiesen et al [61] (2005) determined the force system yielded by different designs of T-loop. The result showed that closing loops made 0.019 x 0.025 inch beta titanium wire gave higher horizontal forces than loops made from 0.017 x 0.025 beta titanium wire. The size of the wire had only a small effect on the M/F ratio generated.

The insertion of 180 degree gable bends in T-loops with and without helices increased the M/F ratio, but had a variable effect on the magnitude of the horizontal force. In general, the beta-titanium T-loops without helices yielded higher magnitudes of horizontal force and M/F ratios than the beta-titanium T-loops with helices.

After insertion of gable bends, plain T-loops and T-loops with helices yielded high M/F ratios, whereas loops without gable bends generated low M/F ratios.

Safavi et al [52] (2006) compared 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 rectangular (0.016 x 0.022 inch) stainless steel wire in a segmented arch with the finite element method (FEM). The highest horizontal and vertical forces were produced by the L-loop (with and without preactivation bends) and in most cases the lowest forces were produced by the VHC loop. Loops with preactivation bends produced marked changes in the M/F ratio and loops without preactivation bends low, but relatively constant, M/F ratios over the full range of activation. Stainless steel opus and T-loops without preactivation bends had constant M/F ratios, but both loops failed to deliver the optimum M/F ratio of 10:1.

Proffit [46] (2007) stated that the two major reasons to extract teeth in orthodontic treatment is 1) to provide space to align the remaining teeth in the presence of severe crowding and 2) to allow teeth to be moved so that protrusion can be reduced or skeletal class 11 or class 111 problems can be camouflaged. To obtain the desired result of closing extraction spaces within the arch, it is essential to control the amount of incisor retraction vs molar-premolar protraction**.**

Lim et al [38] (2008) investigated the temperature effects on the Forces, Moments and Moment to Force Ratio of Nickel-Titanium and TMA symmetrical T-loops. The M:F ratios of NiTi loops were less affected, with no significant changes with temperature for the 15° and 30°preactivation loops, although some change was noted for the non-preactivated loops. TMA wires showed significance for some force measurements, but were generally not influenced by temperature.

Sander et al [54] (2008) reported that in the superelastic range, the force does not increase in proportion to the deflection. This is the principal advantage of nickel titanium archwires: they can be deformed over long distances, while continuing to deliver clinically acceptable force levels.

Nickel titanium wires do not show superelastic behaviour at deflections of less than about 1mm, because such deflections do not adequately exceed the elastic limit of the wire. In this low range, Hooke's law applies: stress is proportional to strain. After heating nickel-titanium wire returns to the condition that was programmed into it during the manufacturing process. This behavior is referred to as "Shape memory''. To program a change in shape memory requires temperatures of 300-520°c, depending on the material property desired.

Martins et al [40] (2009) in a prospective clinical investigation evaluated the movements produced during partial retraction of the maxillary and mandibular canines with a group A TTLS. Based on the result of this study, the M/F ratios typically recommended are excessive and should be different for the posterior and anterior segments. Most laboratory and experimental estimates of MF ratios to produce translation varying from 10mm to 14mm are too high. The differences are due to the Line of force application, which is usually evaluated perpendicular to the teeth and overestimates the resistance offered by the bone. When teeth are initially tipped, the distance between the LFA and the center of resistance becomes smaller than when they are upright.

Rose et al [50] (2009) investigated the loads (forces), moments, and moment-to-force ratios (M:F) generated during the activation and deactivation of closing loops made of rectangular nickel-titanium (NiTi) and titanium-molybdenum alloy (TMA) wires incorporating either 0º, 15º, or 30º of preactivation. To set the shape of the loops in the NiTi wires, they were formed and clamped around stainless steel pins on a template and heated at 510°c for 9 minutes in a crown furnace. The non preactivated closing loops failed to produce an optimum M:F for theoretical tooth movement via translation. TMA generally produced a higher mean force over its activation range compared with the equivalent NiTi closing loops. With increasing degrees of preactivation, the M:F also increased over the deactivation range for all closing-loops. All preactivated TMA and NiTi closing-loop specimens produced an M:F>_ 10:1 at some point in their deactivation range, irrespective of the force delivered. In the assumed optimal biologic force range (50-150g), the NiTi preactivated closing loops produced an M:F of >_10:1 over a greater deactivation range than did their TMA counterparts.

Sander et al [53] (2009) described a method for reprogramming the memory of superelastic nickel titanium wires using a commercially available device, the Memory-Maker. The Memory-Maker permanently changes the conformation of superelastic nickel titanium wires without destroying their superelastic properties, allowing them to be used for a wide variety of specific clinical applications that are not easily addressed with preformed commercial archwires. The wires can be reprogrammed repeatedly as long as they are not overheated. Customized auxiliaries can be constructed and segments of nickel titanium archwires adjusted to achieve desired movements of individual teeth. Exploiting the superelastic properties of nickel titanium wires can help achieve optimal outcomes while reducing costs and treatment times.

Bolender et al [5] (2010) stated that most NiTi archwires did not display any superelasticity in torsion at average oral temperature. The tested braided stainless steel D-Rect rectangular archwire displayed better torsional properties at 35ºC than most NiTi archwires of the same dimensions.

Vojtech et al [63] (2010) stated that the shape setting treatment is generally carried out at moderate temperatures (around 500ºc) and its purpose is to induce relaxation of a material for achievement of a desired stable shape.

Moderate temperatures and short times are used to prevent the permanent deformation of the NiTi specimen and to maintain their superelastic behavior. High resolution SEM image of the wire annealed at 540ºc for 8 minutes revealed the presence of fine precipitates in austenitic matrix. . Annealing temperatures between 410-460ºc improve the strength to some extend. At higher temperatures above 485ºc, the strength reduces.

Gajda et al [25] (2011) experimentally quantified the effects of the loop design on three-dimensional orthodontic load systems of two types of commercial closing loop archwires: Teardrop and Keyhole. An orthodontic force tester and custom-made dentoform were used to measure the load systems produced on two teeth during simulated space closure. The system included three force components along and three moment components about three clinically defined axes on two target teeth: the left maxillary canine and the lateral incisor. The different designs delivered similar loading patterns. The component magnitudes are dependent on the design. All of the designs result in lingual tipping of the teeth, canine lingual-mesial displacement, canine crown-mesial-in rotation, and incisor crown-distal-in rotation.

Alfredo Gilbert [1] (2011) introduced a system for designing and bending archwires more precisely and rapidly, called LAMDA (Lingual Archwire Manufacturing and Design Aid). The LAMDA robot incorporates a heater that can raise the temperature of a nickel titanium archwire to 600ºF, making it possible to bend the wire without losing its capacity to transform reversibly between the austenitic and martensitic phases.

Keng F.Y. et al [33] (2011) conducted a prospective randomized controlled clinical trial to evaluate the rate of space closure and tooth angulation during maxillary canine retraction using preactivated T-loops made from TMA and NiTi wires. The loops were activated 3mm at each visit to deliver a load of approximately 150 grams. There was no difference in the rate of space closure or tooth angulation between preactivated TMA or NiTi T-loops. The NiTi loops possessed a greater ability to retain and return to their original shapes following cyclical activation

MATERIALS AND METHODS

This **in-vitro** study consisted of ninety samples of symmetrically placed T-loops made of two different alloys. Samples were divided into three groups, Group A: TMA, Group B: NiTi (furnace), Group C: NiTi (SMAS). Each group comprised of thirty samples. Ten samples for each degree of preactivation for all three groups were tested. Samples were constructed from commercially available 0.018x0.025 inch austenitic NiTi wire (3M Unitek) and 0.017x0.025 inch TMA wire (Ormco).The dimensions of the T-loop was- height 7mm, width 10mm and radius 2mm.

Preactivation bends of 0°, 15° and 30° were given at the junction of vertical arm and the occlusal horizontal arm of the closing loops on both sides. (Fig.1)

Two methods were used for the fabrication of NiTi T- Loops:

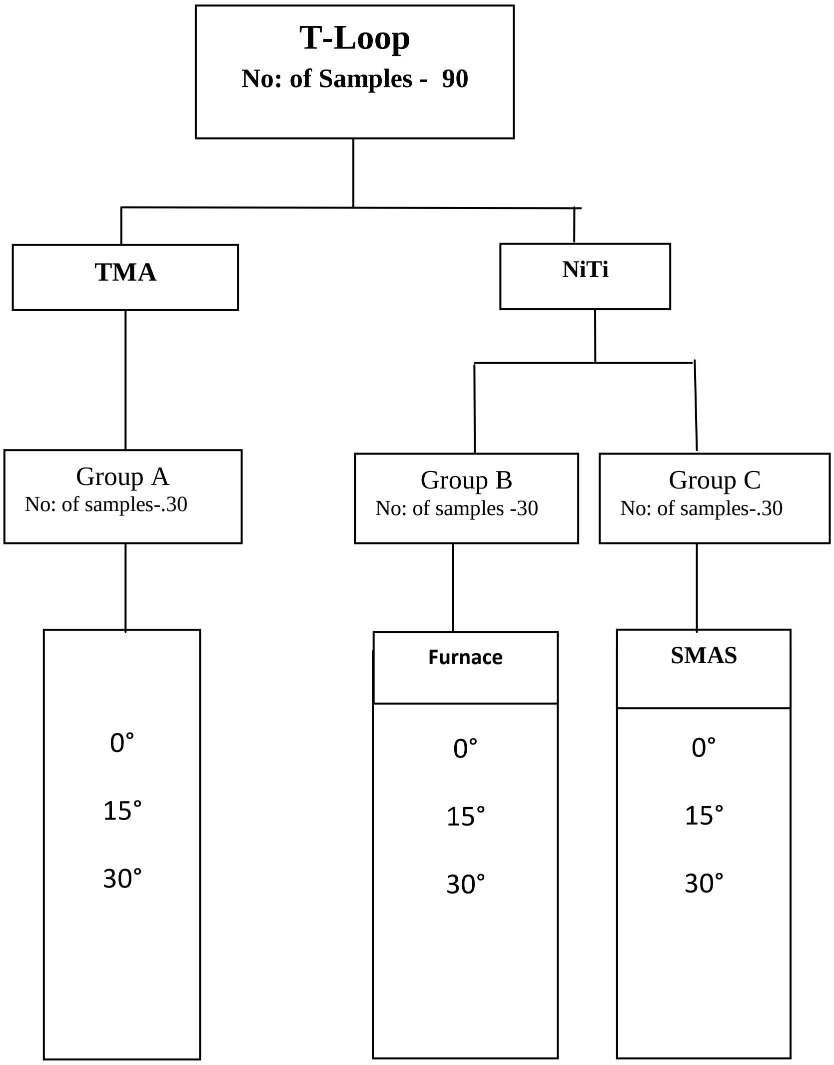
1) After forming the loops with three prong it was incorporated in Orthocal and held in place with ligature wire (Fig.3.) and heated at 510°c for 9 minutes in a muffle furnace (Fig 4)
2) Using SMAS, a Shape memory alloy shaper (Jaypee General Agencies) which is a pulsed-heat induction device. (Fig.5) This device consists of two electrically connected pliers and rectangular foot switch. The wire is grasped with the pliers and electric current is applied.

The measurements were taken for each sample at an interval of 1mm from the neutral position during the activation and deactivation range to maximum of 7 mm for TMA and 8 mm for NiTi closing loops.(Fig 6).

A custom made **loop testing apparatus (**IIT Chennai) was used to measure the forces and moments generated. The apparatus consisted of a force transducer, moment transducer, and a displacement sensor. Two holders on either side of the apparatus were used to hold the ends of the wire samples. (Fig 7&Fig 8)

A stainless steel twin bracket with 0° torque and 0° angulation, slot size of 0.018 inch (Ormco) was positioned onto the lower end of **force transducer** so that upper end (representing the posterior) of the arch wire was in line with the center axis of the load cell. The upper part of the apparatus which consisted of the force transducer, holder and the first bracket was movable. A second bracket was attached 10 mm from the force transducer to which **moment transducer** was connected. The lower end of the wire representing the anterior segment was attached to a third bracket at 6mm from moment transducer and was secured with a fixed holder. Loops were mounted in the brackets so that they were positioned equidistant between the force and moment transducer. Wire samples were held in place using SS ligature wire (Fig 9).

FLOW CHART


T-Loop
No: of Samples - 90
TMA
NiTi
Group A
No: of samples-.30
Group B
No: of samples -30
Group C
No: of samples-.30
Furnace
SMAS
0°
15°
30°
0°
15°
30°
0°
15°
30°

Room temperature at the time of measuring was 32°c ± 0.5°c.

The following assumptions were made:

1. The heat setting of the closing loop formed in the NiTi test wire have a negligible effect on the superelastic properties of the wire i.e, a complete single phase material after loop setting with no change in the transitional temperature range.
2. The interbracket distances were for an average clinical patients.
3. Tightening of the holders had a minimal effect on the wire material.
4. The forces and moments in a single plane had no significant influence on those generated out of the plane.

Statistical analysis

Independent sample T- test (parametric test) was used to identify the statistical difference between M:F ratio generated by T-loops constructed from different wire materials (TMA and NiTi-furnace, TMA and NiTi-SMAS) and between the M/F ratio generated by Niti T-loops fabricated using two different methods - NiTi(furnace) and NiTi (SMAS). Friedman Test (nonparametric test) was used to identify the statistical difference between the M/F ratio generated by T-loops with different preactivations (0º,15º and 30º). P value ≤ 0.01 is considered as highly significant. P value ≤ 0.05 is considered as statistically significant.

Statistical analysis were undertaken using the Statistical package for the Social Sciences (SPSS) Version 15.

Preactivation bends - 0°, 15° and 30°

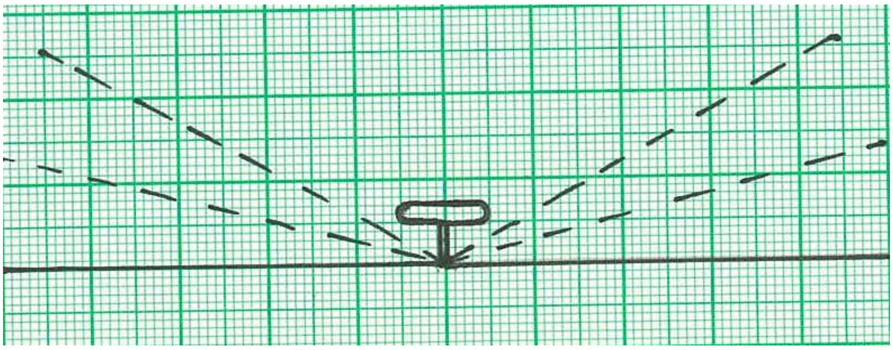


Fig.1.

ARMAMENTARIUM

Fig.2

Template for stabilizing the NiTi (furnace) closing loops

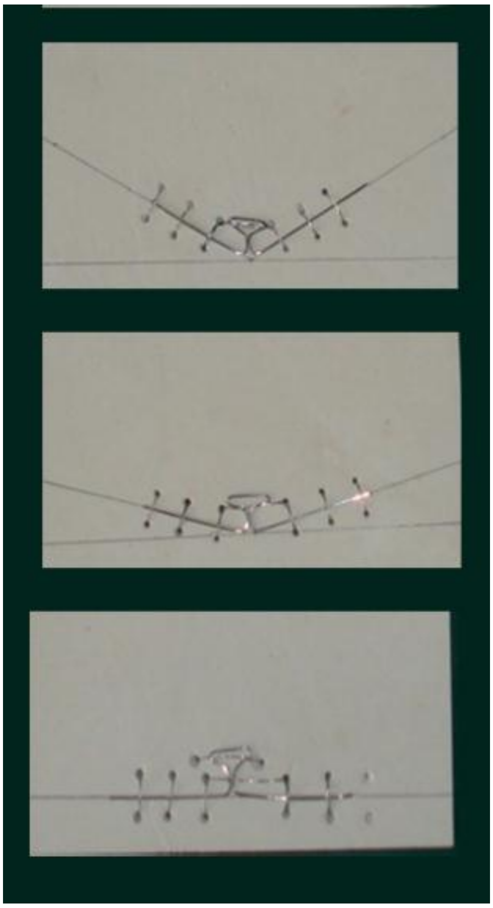


Fig.3

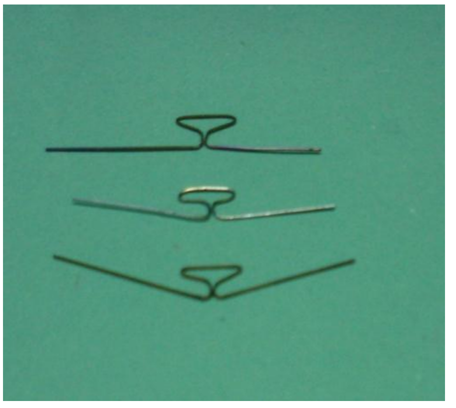
MUFFLE FURNACE

Fig 4.

SMAS

Fig 5

NiTi-(Furnace)



NiTi-(SMAS).

TMA.

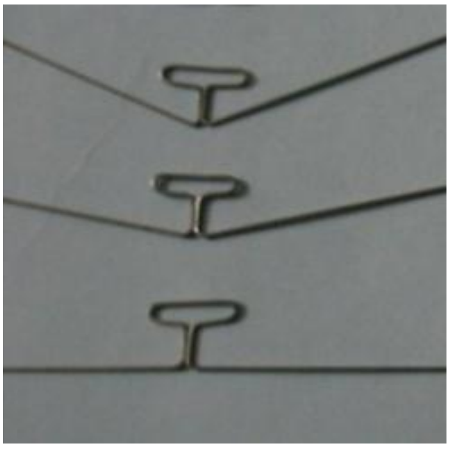


Fig.6

Loop testing apparatus

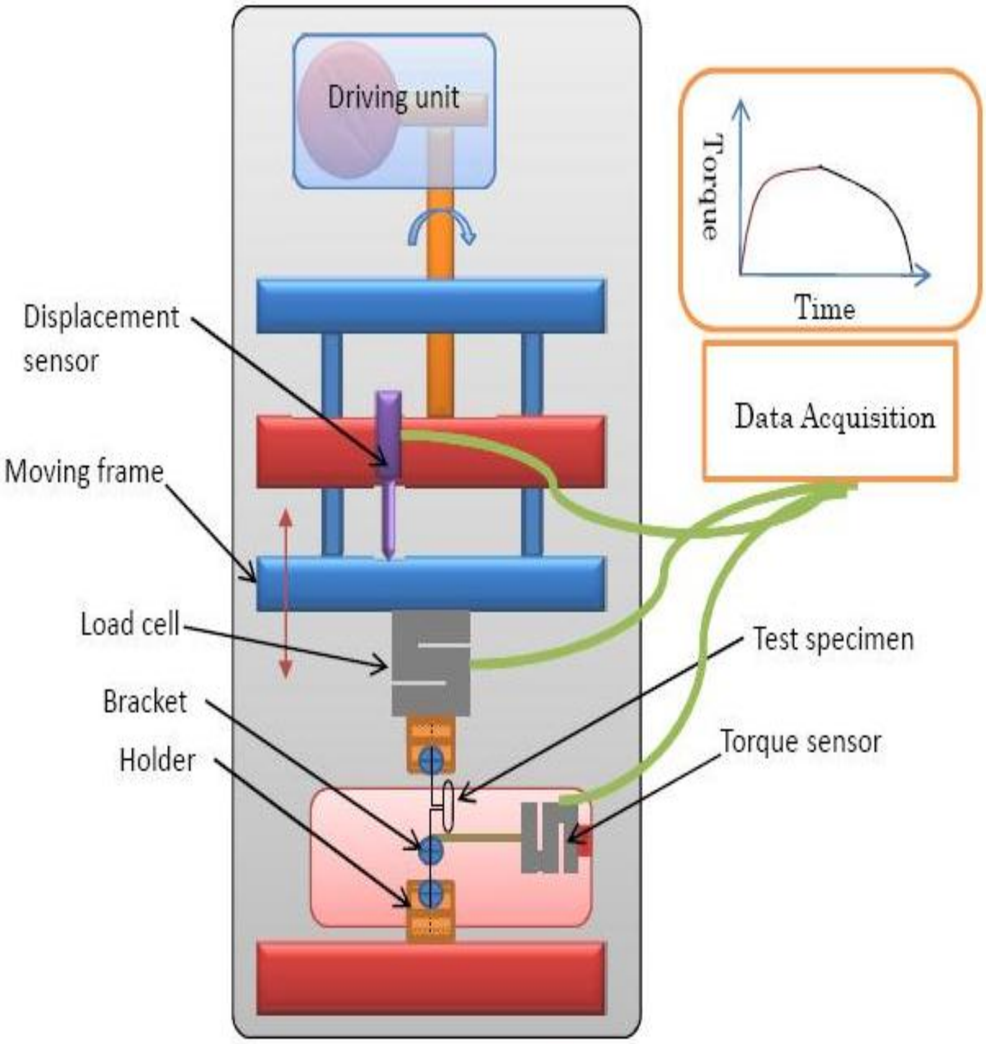


Fig.7

Loop testing apparatus

Fig.8

Fig.9

RESULTS

Comparison between the M:F generated by Group A - TMA and Group B - NiTi (furnace) T-loops showed statistically significant difference between two groups. The difference between the two groups showed significance at 1% level for 15º and 30º. The M:F generated by NiTi (furnace) group was higher than TMA group for 15º and 30º preactivations at all displacements measured. The M:F generated by 0º preactivation showed some variation. **(Table I)**

Comparison between the M:F generated by Group A - TMA and Group C - NiTi (SMAS) T-loops showed statistically significant difference for 15º and 30º at all displacements from 5mm to 1mm. At 6mm displacement only 15º showed statistically significant difference. At 1mm displacement 0º, 15º and 30º preactivation showed statistically significant difference between two groups. The M:F generated by NiTi (SMAS) T-loop was higher than TMA T-loop at all displacements measured except for 0º preactivation. M:F of 0º NiTi (SMAS) was higher than TMA T-loop only at 1mm displacement. **(Table II)**

Comparison of M:F produced by Group B -NiTi (furnace) and Group C - NiTi (SMAS) showed statistically significant difference between two groups for 30º preactivation at all displacement measured. 0º and 15º showed some variations. **(Table III)**

Comparison between the M:F generated by 0º, 15º and 30º preactivated T-loops showed that, with an increase in the degree of preactivation the M:F generated also increases. This was observed in the case of TMA as well as both the NiTi groups. 0º preactivation showed the lowest M:F and 30º preactivation showed the highest. The increase in the M:F with the increase in the degree of preactivation was statistically significant ($P \leq 0.05$) at 5% level. **(Table IV)**

The force levels produced by NiTi T- loops were significantly lower than those of TMA T-loops.. The activation force levels were generally higher than the deactivation force levels for NiTi T- loops.

TABLE I M:F : COMPARISON OF TMA AND NITI (furnace) T- LOOPS. Independent t-test.

DEACTIVATION (mm)	PRE ACTIVATION	TMA	NiTi (furnace)	P value
7	0°	-	5.5±0.24	-
	15°	-	6.43±0.46	-
	30°	-	7.9±0.18	-
6	0°	4.9±0.13	5.3±0.16	0.028 *
	15°	5.5±0.04	7.2±0.23	<0.001 **
	30°	7.1±0.33	9.5±0.07	<0.001 **
5	0°	5.23±0.27	5.1±0.40	0.658
	15°	5.8±0.12	7.7±0.32	0.001 **
	30°	7.3±0.23	10.3±0.21	<0.001 **
4	0°	5.4±0.26	5.0±0.12	0.073
	15°	6.1±0.20	8.1±0.38	0.001 **
	30°	7.7±0.20	11.5±0.31	<0.001 **
3	0°	5.2±0.21	5.07±0.26	0.529
	15°	6.4±0.11	8.6±0.41	0.001 **
	30°	8.3±0.05	12.1±0.08	<0.001 **
2	0°	4.9±0.30	5.4±0.26	0.095
	15°	6.9±0.33	9.0±0.35	0.002 **
	30°	10.5±0.06	13.1±0.15	<0.001 **
1	0°	4.5±0.03	5.07±0.09	0.001 **
	15°	7.2±0.20	9.3±0.35	0.001 **
	30°	11.6±0.10	13.5±0.31	<0.001 **
0	0°	-	-	-
	15°	-	-	-
	30°	-	-	-

Note : 1.** denotes significance at 1% level.

2.* denotes significance at 5% level

TABLE II. M:F : COMPARISON OF TMA AND NITI (SMAS) T- LOOPS. Independent t- test.				
DEACTIVATION (mm)	**PRE ACTIVATION**	**TMA**	**NiTi (SMAS)**	**P value**
7	0°	-	4.9±0.12	-
	15°	-	5.8±0.08	-
	30°	-	6.9±0.13	-
6	0°	4.9±0.13	4.6±0.23	0.121
	15°	5.5±0.04	6.1±0.11	0.001 **
	30°	7.1±0.33	7.5±0.27	0.180
5	0°	5.23±0.27	5.1±0.38	0.647
	15°	5.8±0.12	6.7±0.22	0.003 **
	30°	7.3±0.23	8.6±0.31	0.004 **
4	0°	5.4±0.26	4.8±0.14	0.024 *
	15°	6.1±0.20	7.3±0.08	0.001 **
	30°	7.7±0.20	9.5±0.18	<0.001 **
3	0°	5.2±0.21	4.9±0.21	0.155
	15°	6.4±0.11	8.3±0.15	<0.001 **
	30°	8.3±0.05	10.9±0.22	<0.001 **
2	0°	4.9±0.30	4.7±0.16	0.366
	15°	6.9±0.33	8.63±0.15	0.001 **
	30°	10.5±0.06	11.7±0.16	<0.001 **
1	0°	4.5±0.03	5.1±0.31	0.029 *
	15°	7.2±0.20	9.1±0.14	<0.001 **
	30°	11.6±0.10	12.2±0.11	0.002 **
0	0°	-	-	-
	15°	-	-	-
	30°	-	-	-

Note : 1.** denotes significance at 1% level.

2.* denotes significance at 5% level.

TABLE III. M:F : COMPARISON OF NiTi (furnace) AND NiTi (SMAS) T-LOOPS. Independent t- test.				
DEACTIVATION (mm)	**PRE ACTIVATION**	**NiTi (FURNACE)**	**NiTi (SMAS)**	**P value**
7	0°	5.5±0.24	4.9±0.12	0.018 **
	15°	6.43±0.46	5.8±0.08	0.079
	30°	7.9±0.18	6.9±0.13	0.001 **
6	0°	5.3±0.16	4.6±0.23	0.012 **
	15°	7.2±0.23	6.1±0.11	0.002 **
	30°	9.5±0.07	7.5±0.27	<0.001 **
5	0°	5.1±0.40	5.1±0.38	1.000
	15°	7.7±0.32	6.7±0.22	0.011 **
	30°	10.3±0.21	8.6±0.31	0.001 **
4	0°	5.0±0.12	4.8±0.14	0.133
	15°	8.1±0.38	7.3±0.08	0.023 *
	30°	11.5±0.31	9.5±0.18	0.001 **
3	0°	5.07±0.26	4.9±0.21	0.438
	15°	8.6±0.41	8.3±0.15	0.300
	30°	12.1±0.08	10.9±0.22	0.001 **
2	0°	5.4±0.26	4.7±0.16	0.017 **
	15°	9±0.35	8.63±0.15	0.171
	30°	13.1±0.15	11.7±0.16	<0.001
1	0°	5.07±0.09	5.1±0.31	0.867
	15°	9.3±0.35	9.1±0.14	0.410
	30°	13.5±0.31	12.2±0.11	0.002 **
0	0°	-	-	-
	15°	-	-	-
	30°	-	-	-

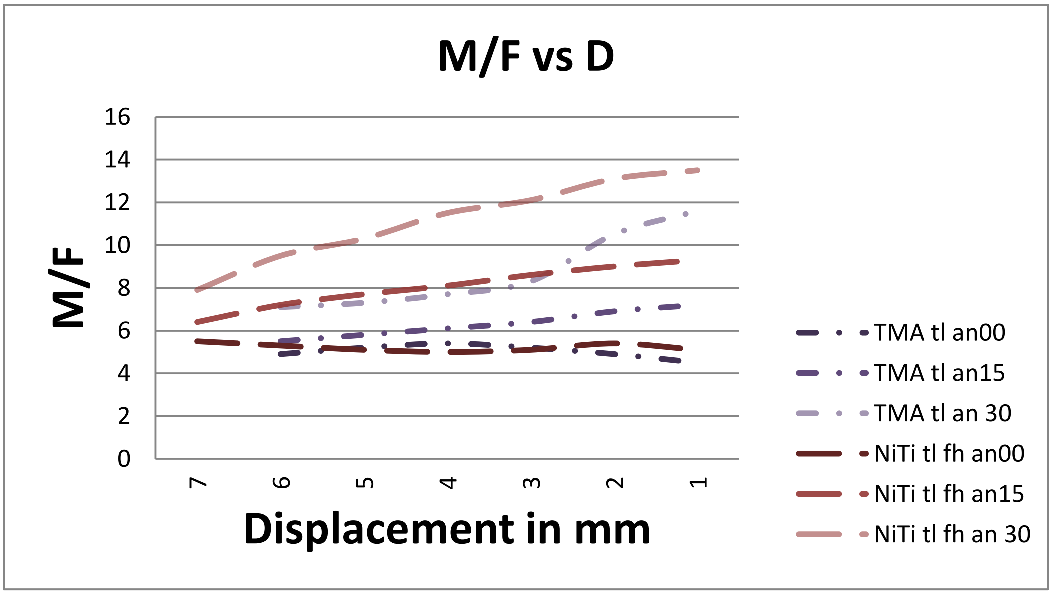
Note : 1.** denotes significance at 1% level.

2.* denotes significance at 5% level.

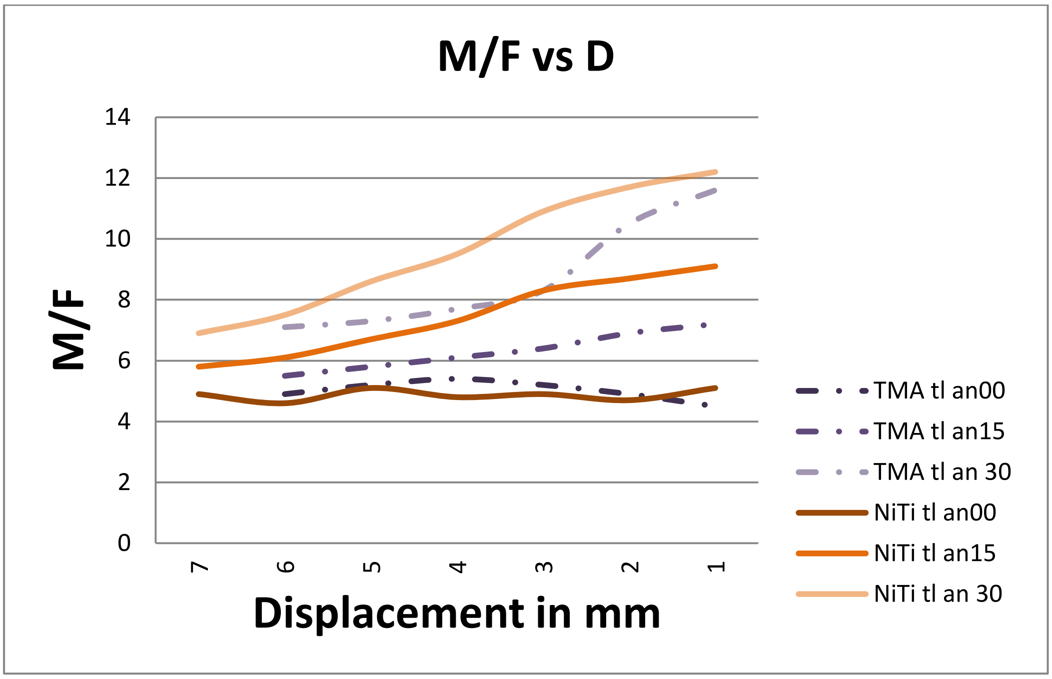
TABLE IV. M:F : COMPARISON OF PREACTIVATIONS WITHIN TMA AND NITI T-LOOPS. Friedman Test.					
DEACTIVATION (mm)	**MATERIAL**	**0°**	**15°**	**30°**	**P value**
7	TMA	-	-	-	-
	Niti(FURNACE)	5.5±0.24	6.43±0.46	7.9±0.18	0.050 *
	NiTi (SMAS)	4.9±0.12	5.8±0.08	6.9±0.13	0.050 *
6	TMA	4.9±0.13	5.5±0.04	7.1±0.33	0.050 *
	NiTi(FURNACE)	5.3±0.16	7.2±0.23	9.5±0.07	0.050 *
	NiTi (SMAS)	4.6±0.23	6.1±0.11	7.5±0.27	0.050 *
5	TMA	5.23±0.27	5.8±0.12	7.3±0.23	0.050 *
	NiTi(FURNACE)	5.1±0.40	7.7±0.32	10.3±0.21	0.050 *
	NiTi(SMAS)	5.1±0.38	6.7±0.22	8.6±0.31	0.050 *
4	TMA	5.4±0.26	6.1±0.20	7.7±0.20	0.050 *
	NiTi(FURNACE)	5±0.12	8.10±0.38	11.5±0.31	0.050 *
	NiTi(SMAS)	4.8±0.14	7.3±0.08	9.5±0.18	0.050 *
3	TMA	5.2±0.21	6.4±0.11	8.3±0.05	0.050 *
	NiTi(FURNACE)	5.07±0.26	8.6±0.41	12.1±0.08	0.050 *
	NiTi(SMAS)	4.9±0.21	8.3±0.15	10.9±0.22	0.050 *
2	TMA	4.9±0.30	6.9±0.33	10.5±0.06	0.050 *
	NiTi(FURNACE)	5.4±0.26	9±0.35	13.1±0.15	0.050 *
	NiTi(SMAS)	4.7±0.16	8.63±0.15	11.7±0.16	0.050 *
1	TMA	4.5±0.03	7.2±0.20	11.6±0.10	0.050 *
	NiTi(FURNACE)	5.07±0.09	9.3±0.35	13.5±0.31	0.050 *
	NiTi(SMAS)	5.1±0.31	9.1±0.14	12.2±0.11	0.050 *
0	TMA	-	-	-	-
	NiTi(FURNACE)	-	-	-	-
	NiTi(SMAS)	-	-	-	-

Note : 1.* denotes significance at 5% level.

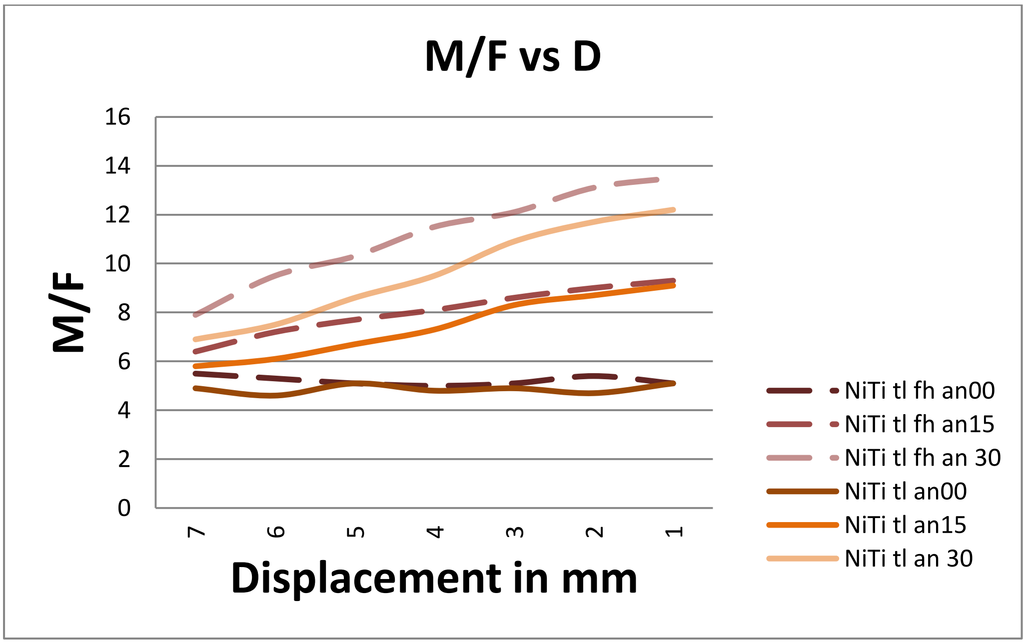
GRAPH I. REPRESENTING THE M/F RATIO OF TMA AND NiTi-FURNACE (NiTi tl fh)T-LOOPS.



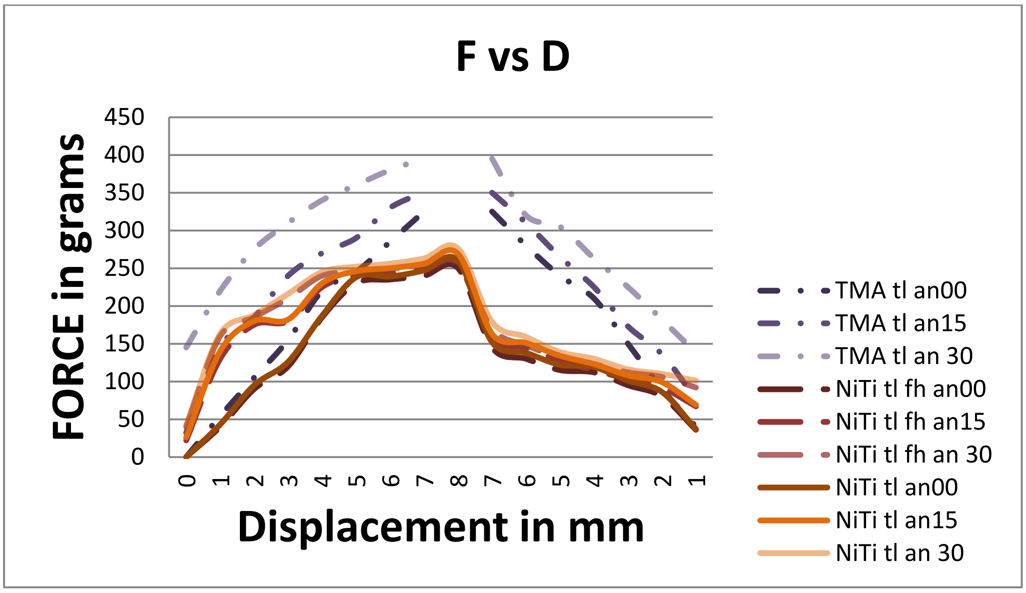
GRAPH II. REPRESENTING THE M/F RATIO OF TMA AND NiTi-SMAS (NiTi tl) T-LOOPS.



GRAPH III. REPRESENTING THE M/F RATIO OF NiTi –FURNACE (NiTi tl fh) AND NiTi–SMAS (NiTi tl) T-LOOPS.



GRAPH IV. REPRESENTINGTHE FORCE VALUES FOR TMA, NiTi-FURNACE (NiTi tl fh) and NiTi –SMAS (NiTi tl) T-LOOPS.



DISCUSSION

The fundamental goal of orthodontic treatment is to move teeth a prescribed distance in a predetermined direction to enhance the esthetic and functional aspects of occlusion and achieve a stable result.[41]

The orthodontic appliance is the clinician's primary tool for initiating and sustaining the biological processes that control tooth movement. This is achieved by applying a force system that displaces a tooth, or segment of teeth, causing stresses in the periodontal ligament which produce physical, chemical, and electrical signals that are sent to the surrounding cells and tissues. Both the quantity and quality of tooth displacement can be altered by varying the magnitude and direction of the moments and forces applied mechanically to the teeth. Understanding the relationship between appliance activation and the resultant stresses produced at the level of the periodontal ligament is the first step toward understanding the biological mechanisms that allow clinicians to move teeth predictably.

From a cellular point of view, distribution of stress (force per unit area), distortion of the periodontal ligament (shear stress, strain), and bone deformation (strain) are critical factors, and the remodeling response is directly related to stress and strain levels within the periodontium. The orthodontic force as an extrinsic mechanical stimulus evokes a biologic cellular response that aims to restore equilibrium by remodeling of the periodontal supporting tissues. Because of difficulty to measure stress and strains within the periodontal ligament of loaded teeth directly, measuring the forces that are applied directly to teeth with known root surface areas can provide an estimate of these parameters.

Schwarz [61] proposed the classic concept of the optimal force. He defined optimal continuous force as “the force leading to a change in tissue pressure that approximated the capillary vessels’ blood pressure, thus preventing their occlusion in the compressed periodontal ligament.’’ According to Schwarz, forces well below the optimal level cause no reaction in the periodontal ligament. Forces exceeding the optimal level would lead to areas of tissue necrosis, preventing frontal bone resorption. Tooth movement would thus be delayed until undermining resorption had eliminated the necrotic tissue obstacle.

Schwarz’s definition was slightly modified by **Oppenheim**,[47] who advocated the use of the lightest force capable of bringing about tooth movement, and by **Reitan**,[51] who demonstrated cell-free compressed areas within the pressure site even in cases where light forces were applied and also advocated the use of very light forces.

The current concept of optimal force is based on the hypothesis that a force of a certain magnitude and temporal characteristics (continuous vs intermittent, constant vs declining, etc) would be capable of producing a maximum rate of tooth movement without tissue damage and with maximum patient comfort. The optimal force for tooth movement may differ for each tooth and for each individual patient.[49]

Knowledge of the mechanics required to achieve specific treatment goals is necessary for efficient correction of the malocclusion. Selecting the appropriate force system provides the orthodontist with better control of the tooth movement. Three important variables which are under the control of the orthodontist are the moment-to force ratio, the magnitude of the force and/or moment, and the force constancy. The moment-to-force ratio determines the center of rotation of a tooth or segment of teeth, thus allowing translation, tipping or root movement. An optimal force magnitude can be found which will rapidly move the teeth while minimizing patient pain or discomfort and having little or no tissue damage. Force constancy refers to the maintenance of the desired force level throughout the orthodontic tooth movement. Consideration of each of these variables allows the clinician to specifically adjust the orthodontic appliance to obtain the desired tooth movement.

According to **Smith and Burstone,** [63] forces produce translation, 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. The tendency to rotate is due to the moment of the force, which is equal to force magnitude multiplied by the perpendicular distance from the point of application of force to the center of resistance. 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. A moment-to-force ratio of about 8:1 produce controlled tipping, 10:1 results in translation of tooth and 12:1 produce root movement.

Rudolph DJ et al, [55] based on 3-dimensional finite element model concluded that the principal stress from a tipping force was located at the alveolar crest. For bodily movement, stress was distributed throughout the periodontal ligament.

Closure of extraction spaces is an integral stage of many orthodontic treatment plans. In goal oriented orthodontics the closure of these spaces requires an understanding of the mechanical system utilized.

The two major reasons to extract teeth in orthodontic treatment is 1) to provide space to align the remaining teeth in the presence of severe crowding and 2) to allow teeth to be moved so that protrusion can be reduced or skeletal class 11 or class 111 problems can be camouflaged.[49]

Space closing mechanics can be broadly divided into two categories, friction (sliding) and frictionless (loop) mechanics.

In sliding mechanics, the wire and position of the bracket give control of tooth movement, whereas in a loop-spring system, control is built into the spring. The sliding mechanisms in any application other than simple tipping movements have two disadvantages: (1) The friction may stop tooth movement entirely as one approaches translatory types of movement. (2) Force magnitudes cannot be easily determined since the amount of friction is relatively unknown and unpredictable. It is difficult to precisely control the force level applied for retraction which leads to tipping, binding of archwire, lack of vertical control, risk of anchor loss and incisor extrusion. These problems can be overcome by using loop or frictionless mechanism.

The major advantages of frictionless mechanics is 1) the absence of friction between bracket and wire, 2) a known force system is delivered to teeth because there is no dissipation of force by friction and 3) M:F ratio is predictable and controllable during retraction. [16]

Hence the **aim of this study** is to evaluate the M:F ratio of T Loops made of TMA and NiTi.

A variety of vertical loop configurations are used by clinicians to close spaces between individual teeth or groups of teeth. Although it may be possible to design loops or retraction springs to deliver an adequate moment-to-force ratio for controlled tipping around the apex of an incisor or a canine, translatory movements are not possible, considering the intraoral limitations on spring height. This can be overcome by the placement of gable bends or angulation in a loop or retraction spring.

If a higher moment-to-force ratio is required, it is necessary to increase the vertical height of the loop. Additional horizontal wire placed gingivally will also raise the moment-to-force ratio for any given loop length. [16]

Faulkner et al [24] reported that typical stainless steel vertical loop has two major limitations. First, its activation range is very restricted; second, the M:F ratio produced is also well below ideal if controlled tipping or translation is desired.

One of the common space closure appliances is the "T-loop". In the Segmented Arch Technique, developed by **Burstone** in 1962, utilizes T-loop space closure springs for anterior retraction, symmetric space closure, or posterior protraction. These springs can be preactivated to deliver the required counter moments as they are deactivated. These counter moments should be the same magnitude as the tipping moments to achieve bodily tooth movement. The basic T-loop design uses 0.017x 0.025 inch TMA wire. The segmental T-loop space closure principles can also be applied to space closure on a continuous arch.

Raboud et al [50] reported that T loops produce less forces than the vertical loops. For TMA T spring designs, the use of more material, coupled with the different material properties of the TMA, results in force systems closer to those necessary for translation. The elastic activations possible are also larger than with the stainless steel vertical loops.

Burstone[13] reported that a typical vertical loop does not deliver high enough moment-to-force ratios for a translation. The rapid change in moment-to-force ratio with the vertical loop is problematic because biologically it is not desirable to keep changing areas of stress in the periodontal ligament.

Faulkner et al [24] reported that TMA is superior to stainless steel for the appliance design because of a large ratio of yield stress to the elastic modulus.

Based on these factors, **in this study** we have used a T loop design and used TMA instead of stainless steel as one of the material tested. We have tested a symmetric T loop design. On outline this **in vitro study** comprised of **three** groups- Group: A- TMA, Group: B- NITI (furnace), Group: C-NITI (SMAS). Thirty samples were tested for each group.

Kuhlberg and Burstone [36] studied the effects of off-center positioning on the force system produced by symmetric T-loop springs. He concluded that a centered T-loop produces equal and opposite moments with negligible vertical forces. 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. A standard shaped T-loop can be used for differential anchorage requirements by altering the activation and mesial-distal position of the spring.

Ronay et al [53] reported that the interbracket position of the V bend is crucial for the force system delivered to the two teeth, be they two adjacent teeth or, as in the segmented technique, units of teeth with a larger interbracket distance. If the interbracket distance is divided into thirds and if the position of the V bend is in the central third, it will provide two equal and opposite moment. With increasing eccentricity, the moment on the tooth closest to the bend will be bigger and smaller at the distant tooth.

Recent advances in orthodontic wire alloys have resulted in a varied array of wires that exhibit a wide spectrum of properties.

Kusy [37] gave the following criteria for an ideal archwire -esthetic, good range, tough, poor biohostability, good springback, low friction, weldable, springy, formable, biocompatible, resilient and strong. He stated that specific wire will do some things well and others poorly, but no wire will do it all.

Historically, gold alloy wires were used in orthodontic practice but have minimal use currently because of their much greater cost compared to the popular base metal wires. By the 1950's Stainless steel alloys and Cobalt-Chromium-Nickel alloys were used for orthodontic wires.[7]

Beta-titanium alloy was conceived for orthodontic use by **Burstone and Goldberg**.[14] Beta titanium has a modulus of elasticity that is less than that of stainless steel and about twice that of nitinol. This makes its use ideal in situations in which forces less than those of stainless steel are necessary and in instances in which a lower modulus material such as nitinol is inadequate to produce the desired force magnitudes. The springback for Beta-titanium is superior to that of stainless steel and can be deflected almost twice as much as stainless steel wire without permanent deformation. Beta-titanium wires also deliver about half the amount of force as do comparable stainless steel wires.

The first nickel-titanium orthodontic alloy known as Nitinol was introduced to the profession by **Andreasen** and **Hillman** in 1971 based on the original research of **Buehler.** It has an excellent springback property, low stiffness and has a unique property called shape memory.

Shape memory is a phenomenon occurring in the alloy that is soft and readily amenable to change in shape at a low temperature, but can easily be reformed to its original configuration when it is heated to a suitable transition temperature. The shape memory effect is associated with a reversible martensite-austenite transformation, which occurs rapidly by crystallographic twinning at the atomic level.[7] The temperature range for the start and completion of the transformation to that particular structure is referred to as **Transition Temperature Range.**

One- way effect can be achieved by placing a bend in the NiTi wire while it is in the cold martensitic phase. During subsequent heating of the material beyond the austenite start temperature, the wire is transformed into the SIM phase and the bend disappears completely. Cooling the archwire does not recover the bend. The **two-way effect** is achieved by exposing a nickel titanium wire to a sharp deformation while the wire is in its cold martensitic condition. The martensitic deformation is partially reversed when the wire is warmed and when the wire is cooled it partially recovers its original deformation. **All-around effect** can be achieved by heating an archwire in the martensitic phase to 300-520ºc, and a new shape can be programmed into it.[59]

Nickel-titanium wires have a larger recoverable stored energy than stainless steel or beta-titanium wires when activated to the same amount of bending or torsion.[23] Higher springback values provide the ability to apply large activations with a resultant increase in working time of the appliance. This, in turn, implies that fewer arch wire changes or adjustments will be required.[32]

Many new nickel-titanium orthodontic wire alloys have been introduced, and some of these new brands possess the property of **superelasticity.** These wires are referred to as A-NiTi, with an active austenitic grain structure.

The super-elastic property is a phenomenon by which the stress value remains fairly constant up to a certain point of wire deformation. At the same time, when the wire deformation rebounds, the stress value again remains fairly constant. This property for A-NiTi wire occurs because of a phase transition in grain structure from austenite to martensite, in response not to a temperature change but to applied force.[49] The phase transformation occurs from the body-centered cubic austenitic form to the hexagonal close-packed martensitic form of NiTi when the stress reaches a certain level during activation. Upon deactivation, the reverse-phase transformation from the martensitic to the austenitic structure takes place when the stress is decreased to an appropriate level.

Burstone et al [18] reported that the **Chinese NiTi** wire developed by **Dr. Tien Cheng** and associates in China delivers a relatively constant force over a long range of action.

Miura et al [45] reported that the **Japanese NiTi** alloy developed by Furukawa Electric Co., Ltd. of Japan possesses excellent springback property, shape memory, and super-elasticity.

Superelasticity is especially desirable because it delivers a relatively constant force for a long period of time, which is considered a physiologically desirable force for tooth movement.

Neo Sentalloy introduced in early 1990's, with true shape memory, has essentially a completely austenitic structure at oral temperature. Copper NiTi wire which is available in temperature variants of 27ºc, 35ºc and 40ºc would be useful at different oral temperatures.

Based on the clinical significance of austenitic NiTi, **in this study** we have used T-loop constructed from TMA and superelastic NiTi wires incorporated with preactivation bends of 0º, 15º and 30º.

Clinical disadvantages of the nickel-titanium orthodontic alloys are that permanent bends cannot readily be placed in the wires and that the wires cannot be soldered. Heat treatment of these wires can be performed by an electrical resistance technique, which allows the orthodontist to prepare custom superelastic nickel-titanium archwires for specific patients. The electrical resistance heat treatment has been exploited to produce archwires for which the level of biomechanical force varies with position (Bioforce**).** [7]

Fabrication of NiTi closing loops

Bourauel et al [6]evaluated superelastic retraction T-loop springs made from Ormco and GAC Sentalloy of different batches by using the Orthodontic measurement and simulation system. Within Ormco or Sentalloy wires, each batch displayed different superelastic behaviour.

Hence, **in our study** the same batch of austenitic NiTi 0.018x0.025 inch archwire of 3M Unitek was used. T-loops were formed using Furnace heating method and pulsed heat induction method (SMAS).

The wire was contoured with three-prong plier. When the wire was bend it underwent a stress induced plastic deformation. Since it crossed the Martensite start (Ms) it did not recover when the stress was removed. The samples were fixed on a template made of Orthocal and held tightly using ligature wire. To induce the shape memory phenomenon of the spring and to incorporate superelasticity, SIM Training was done by incorporating it in Orthocal and maintaining it for 30 minutes allowing the Orthocal to set. Then it was heat treated in **Muffle furnace** at 510^{0}c for 9 minutes.

Rose et al [54] used similar temperature range to set the shape of loops in the NiTi wire. According to **Sander et al,** [58] to program a change in shape memory requires temperatures of 300-520ºc, depending on the material property desired. Studies [35,45] have shown that heat treatment at 600^{0}c resulted in loss of superelasticity.

When the wire was heated it crossed Austenite finish (Af) and the shape was retained since it was maintained in the Orthocal which prevented it from attaining its parent shape, thereby memorizing the new shape.

According to **Vojtech et al** [69] the shape setting treatment is generally carried out at moderate temperatures (around 500ºc) and its purpose is to induce relaxation of a material for achievement of a desired stable shape. Moderate temperatures and short times are used to prevent the permanent deformation of the NiTi specimen and to maintain their superelastic behavior. High resolution SEM image of the wire annealed at 540ºc for 8 minutes revealed the presence of fine precipitates in austenitic matrix.

NiTi loops were also formed with a **pulsed heat induction method** using Shape memory alloy shaper (SMAS). **Sander et al** [57] have demonstrated the use of Memory-Maker, a pulsed heat induction method, to customize nickel titanium archwires. The Memory-Maker permanently changes the conformation of superelastic nickel titanium wires without destroying their superelastic properties. The wires can be reprogrammed repeatedly as long as they are not overheated (Golden yellow colour – workable range, Bluish discolouration – overheated).

A custom **made loop testing apparatus** was used **in this study** to measure forces and moments. Similar method was used by **Rose et al** [54]and **Lim et al** [40] in their studies.

A stainless steel twin bracket with 0° torque and 0° angulation, slot size of 0.018 inch and superelastic (austenitic) NiTi wires (0.018 x 0.025 inch) and TMA wires (0.017 x 0.025 inch) was used **in our study**.

Bolender et al [5] stated that most NiTi archwires did not display any superelasticity in torsion at average oral temperature. The tested braided stainless steel D-Rect rectangular archwire displayed better torsional properties at 35ºC than most NiTi archwires of the same dimensions.

Hence, we have used a 0.018 x 0.025 inch (full size) austenitic NiTi wire in 0.018 inch bracket slot **in our study**.

Interpretation of the Results

The results of our present study shows that with increasing degrees of preactivation, the M:F ratio also increased over the deactivation range for all the T-loops. Hence the **first hypothesis** is accepted. The non preactivated closing loops failed to produce an optimum M:F ratio for theoretical tooth movement via translation.

TMA generally produced a higher force over its activation and deactivation range compared with equivalent NiTi T- loop specimens. These findings are in agreement with those of **Rose et al .**[54]

Burstone and Koenig [16] have reported that the M:F ratio resulting from activating various loop configurations are insufficient to prevent uncontrolled tipping unless Gable bends are included. By placing additional wire apically, the magnitude of the moment increased with respect to the force. But this was not sufficient to give a high enough moment-to-force ratio for translation.

Burstone [13] reported that a TMA T-loop by itself will have a relatively low load-deflection rate and a maximum springback. But this is not sufficient to give a high enough moment-to-force ratio for translation, and a large residual moment must be placed in the spring. Moment-to-force ratio increases during deactivation, but this change is more gradual and hence biologically sounder and more controllable.

The present study showed that the force levels produced by NiTi T-loops were significantly lower than those of TMA T-loops. Superelastic NiTi wires have lower forces because of phase transformations induced by either temperature or mechanical stress.

In the **present study** greater activation range was possible with NiTi T-loops compared to the TMA T-loops.

It is recognized from numerous human studies that light forces (less than 200g) engender adequate biologic responses in the periodontium and that heavier forces are associated with hyalinization of periodontal ligament, undermining bone resorption and are implicated in root resorption.[10]

Based on current and historical literature, the lower deactivation force range of 50g to 150g was accepted in this study as clinically useful values to obtain tooth movement.

Evaluation of the **M:F ratio produced** and the **deactivation range** for various loops that delivered a force of 50 – 150g showed that Group A - TMA with a preactivation of 30º ,Group B - NiTi (furnace) with a preactivation of 30º and Group C - NiTi (SMAS) with a preactivation of 30º produced M:F ratio that exceeded 10:1 at some point during clinically useful deactivation range.

The **M:F** ratio above 10:1 was generated only at 2 mm to 1 mm of deactivation range for Group A – TMA with a preactivation of 30º, 5 mm to 1mm for Group B – NiTi (furnace) with a preactivation of 30º and 3 mm to 1 mm for Group C – NiTi (SMAS) with a preactivation of 30º.

Thus, NiTi produced M:F ratio which is sufficient to produce bodily movement of tooth over a wide range of deactivation with biologically acceptable force levels.

There is a statistical significant difference in moment-to-force ratio between TMA and NiTi loops. Hence the **second hypothesis** is accepted.

The moment-to-force ratio between the Group B - NiTi (furnace) and Group C - NiTi (SMAS) also showed statistically significant difference. Hence the **third hypothesis** is rejected.

Drawbacks of this study-

- Being an in-vitro study the results of this study do not necessarily reflect the behavior of the wires under clinical conditions.
- The ambient temperature at which the measurements were taken were slightly below the oral temperature.
- Direct comparisons were made between the archwires without considering the degree of play between the archwire and bracket.

Problems encountered during this study-

- fabrication of NiTi spring was time consuming and laborious.
- precise bending of NiTi wire to the specification was difficult.

Further scope of the study

There are only very few published information on preactivated NiTi closing loops. Trials are needed to ascertain whether this in-vitro model will respond in a manner consistent with force system stimulus in-vivo. Changes in the mechanical properties of the NiTi wire that might occur during the fabrication of NiTi T-loops also require further investigation.

SUMMARY AND CONCLUSION

This **in-vitro** study evaluated the Moment-to-Force characteristics of symmetrical T-loops with different preactivations, constructed from superelastic NiTi wires and compared it with those of TMA wires. The study consisted of ninety samples of symmetrically placed T-loops made of two different alloys. Samples were divided into three groups, Group A: TMA, Group B: NiTi (furnace), Group C: NiTi (SMAS). Each group comprised of thirty samples. Ten samples for each degree of preactivation for all three groups were tested. Samples were constructed from commercially available 0.018x0.025 inch austenitic NiTi wire (3M Unitek) and 0.017x0.025 inch TMA wire (Ormco). The T-loop used had the following dimensions - height 7mm, width 10 mm and radius 2mm.

Preactivation bends of 0°, 15° and 30° were given at the junction of the vertical arm and the occlusal horizontal arm of the closing loops on both sides.

Two methods were used for the fabrication of NiTi T - Loops:

1) After forming the loop, it was incorporated in Orthocal and heated at 510°c for 9 minutes in a **muffle furnace**.
2) Using SMAS, a Shape memory alloy shaper (Jaypee General Agencies) which is a **pulsed-heat induction device**.

The measurements were taken for each sample at an interval of 1mm from the neutral position during the activation and deactivation range to maximum of 7 mm for TMA and 8 mm for NiTi closing loops.

A custom made **loop testing apparatus (**IIT Chennai) was used to measure the forces and moments generated.

The following conclusions were drawn:

1. The non preactivated T- loops failed to produce an optimum M:F for theoretical tooth movement via translation.
2. With increasing degrees of preactivation, the M:F also increased over the deactivation range for all the T-loops.
3. The M:F generated by NITi loops were greater than TMA T-loops over the deactivation range.
4. NiTi T-loop produced lower force levels which is considered as biologically acceptable, over its activation and deactivation range compared with the equivalent TMA T-loop specimens.
5. NiTi T-loops allowed greater range of activation and the M:F generated was high enough to produce translation and later root movement over a greater range of deactivation. This can be advantageous in the clinical scenario as in a premolar extraction case where complete space closure can be achieved with minimum adjustments.

6. Eventhough there is significant difference in the M:F generated by the two NiTi groups, both the groups provide sufficient M:F for bodily movement of teeth if appropriate preactivations are given. SMAS can be further modified to simplify the fabrication of NiTi T-loops and makes chair side adjustment of the NiTi closing loops possible.

   Further studies are required to confirm these findings.

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