EVALUATION OF TORQUE EXPRESSION OF

FOUR COMMERCIALLY AVAILABLE

SELF-LIGATING BRACKETS –

A FINITE ELEMENT STUDY

Dissertation submitted to

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CERTIFICATE

This is to certify that this dissertation titled **"EVALUATION OF TORQUE EXPRESSION OF FOUR COMMERCIALLY AVAILABLE SELF-LIGATING BRACKETS – A FINITE ELEMENT STUDY"** is a bonafide record of work done by **Dr. AYUSH SHARMA** under my guidance during his postgraduate study period between 2009–2012.

This dissertation is submitted to **THE TAMIL NADU Dr. M.G.R. MEDICAL UNIVERSITY**, in partial fulfillment for the degree of **MASTER OF DENTAL SURGERY IN BRANCH V** – Orthodontics and Dentofacial Orthopedics.

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

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CONTENTS

INTRODUCTION

The term “torque” has two different but related meanings for the orthodontist. On one hand it refers to the bucco-palatal root inclination, which can be measured on the lateral headfilm as the incisor inclination to the anterior cranial base or the maxillary plane, while on the other it describes the activation generated by torsion of the archwire in the bracket slot.[88]

Correct buccolingual inclination of anterior teeth is considered essential for providing good occlusal relationships in orthodontic treatment. Inclination of the maxillary anterior teeth is particularly critical in establishing an esthetic smile line, proper anterior guidance, and a Class I canine and molar relationship[5].

Orthodontist’s define torque around the dental arch such that the x-axis follows the curve of the arch. Torque, in this sense, would be rotation perpendicular to the long axis of the tooth. This could be generated by a rotation through a moment or couple of forces.[4]

The completely programmed bracket system created by **Andrews (1989)**[3], was designed with the objective of using arches without bends. However, in spite of incorporating ideal torque characteristics in the structure of such brackets, in some cases it is necessary to apply additional or individual torque on some teeth. This would be necessary due to several factors[38]**:** mechanical side-effects such as variations in bracket slot and archwire dimension[113], morphological differences in the buccal faces of teeth,[20, 74, 75,111] changes in the position of the brackets, [71,119] different methods of bracket manufacturing[40,121] and orthodontic wires,[18,94] the play between the wire and the bracket slot[13,26], variations in the bracket designs[28], properties of the materials constituting the brackets[39,44,92] and wires[94] and differences between the value of the torque informed by the manufacturer and the real value of the torque in the bracket base[38].

Self-ligating brackets introduced by **Dr. Jacob Stolzenberg (1935)**[98] are ligature-less bracket systems that have a mechanical device built into the bracket to close off the edgewise slot. They are generally smoother for the patients because of the absence of wire ligature and also do not require as much chair time.[9,11,41] The precision arm or the sliding fourth wall accurately locks the archwire within the dimensions of the slot providing robust ligation and controlled tooth movement.

The proclaimed chief advantages of self-ligating systems over conventional appliances include, (a) decrease in treatment duration,[57],[89,] (b) anchorage conservation,[109] (c) asepsis,[28] (d) patient comfort.[33,36,97]

Self-ligating brackets are broadly classified into Active and Passive self-ligating brackets;

1) Active systems - those that have a spring clip that presses against the archwire, such as the InOvation-R (GAC Intl, NY), Time™ (American Orthodontics, USA)
2) Passive systems-those in which the self-ligating clip does not press against the wire such as Damon 3MX (Ormc, USA), SmartClip-3 (3M Unitek,USA).

The finite element method (FEM) is a powerful computer simulation tool, which has been successfully applied to the mechanical study of stress and strain and solving problems in the mechanics of solids and structures.[44,53] This makes it practical to elucidate the biomechanical components such as displacements, stress and strain included in the living structures from various external forces.

In the finite element method, the entire region of the structure is divided into a set of elements that are connected by points called nodes.[105] Element types are decided and each element is assigned its material properties (Young’s Modulus and Poison’s Ratio). The forces and boundary conditions are defined to stimulate loads and constraint of the structures. The structural response is computed and then presented for display.

The FEM has some distinct advantages over other methods of stress analysis.

1) Compared to classical analytical methods, it is able to model much more closely structures of irregular geometries and non-homogeneous or anisotropic material properties and overcomes difficulties inherent in conventional experimental methods.[46]
2) FEM has the potential for the equivalent mathematic modeling of a real object of complicated shape with different material properties. Thus FEM offers an ideal method of accurate modeling of tooth-periodontium system with its complicated 3 Dimensional geometry.[119]
3) The force systems that are used in an orthodontic patient can be complicated, FEM makes it possible to analytically apply various force systems at any point and in any direction.[48]

Clinically, torque control is often required in the maxillary incisors for an ideal inter-incisal angle, adequate incisor contact, and sagittal adjustment of the dentition in order to achieve an ideal occlusion.[5]

Although the self-ligating edgewise bracket was introduced to orthodontists 75 years ago, recent advances in bracket technology have resulted in a number of new self-ligating bracket systems and greater interest in their use. Much of this interest is in response to information comparing the benefits of self-ligating systems with conventional edgewise brackets. Often, this information comes from marketing materials and non-refereed sources claiming that self-ligating bracket systems provide superior treatment efficiency and efficacy.[22]

Because of the complexity of the experimental configuration, only a handful of experimental studies have been presented upon torque expression until now, moreover numerical analyses have not been carried out for torque expression in different self-ligating brackets on the tooth and its supporting structures.[1,24,35,43]

Therefore the aim of the present study was to investigate the torque expression of different self-ligating brackets (active and passive) with various archwire combinations on the tooth and its supporting structures using finite element method.

REVIEW OF LITERATURE

Torque can be defined from a mechanical or from a clinical point of view. Mechanically, it refers to the twisting of a structure about its longitudinal axis, resulting in an angle of twist. Torque is a shear-based moment that causes rotation. Clinically, in orthodontics, it represents the buccopalatal crown/root inclination of a tooth, and it is an orthodontic adaptation used to describe rotation around an x-axis. When applied in an orthodontic archwire/bracket interaction, it describes the activation generated by twisting an archwire in a bracket slot[2].

Depending on magnitude of torsion, the stiffness or resilience of the wire cross section, wire size, edge bevel and manufacturer tolerance, bracket slot size and manufacturer tolerance, engagement angle of the wire in the bracket slot, experimental measurement technique, bracket placement as related to tooth morphology[8,29] and inclination of the tooth, the archwire moves the root of a tooth through the alveolar bone via localized pressure and tension generated by torsion in the archwire[5].

Considering the above factors the review of literature for this study is categorized into two groups:-

1) FEM studies in orthodontics and,

2) Torque and Self Ligating brackets in orthodontics

1) FEM STUDIES IN ORTHODONTICS

Tanne et al (1987)[100] investigated the stress levels induced in the periodontal tissue by orthodontic forces using the three-dimensional finite element method and concluded that during tipping movement, stresses non-uniformly varied with a large difference from the cervix to the apex of the root.

Tanne et al (1988)[104] investigated the relationship between moment to force (M/F) ratios and the centers of rotation by use of the finite element method (FEM). They concluded that the center of resistance was located at 0.24 times the root length measured apical to the level of alveolar crest. The centers of rotation varied with the M/F ratios following a curve of hyperbola. The M/F ratio was -9.53 for root movement (C, at the incisal edge), - 8.39 for translation, and -6.52 for tipping around the apex. It was found that even a small difference in the M/F ratios produced clinically significant changes in the centers of rotation.

Tanne et al (1991)[106] investigated the nature of initial tooth displacements associated with varying root lengths and alveolar bone heights. The results showed that moment-to-force values at the bracket level for translation of a tooth decreased with shorter root length and increased with lower alveolar bone height. In addition, apico-gingival levels of the center of resistance shifted more gingivally to the cervix, or the alveolar crest with a shorter root. However, the relative distances of the centers of rotation from the alveolar crest in comparison with the alveolar bone heights were constant at 0.4 mm, with variations in the root length and alveolar bone height. Because this study showed that root length and alveolar bone height affect the patterns of initial tooth displacements both in the center of resistance and the centers of rotation and also in the amount of displacement, forces applied during orthodontic treatment should take into consideration the anatomic variations in the root length and alveolar bone height so as to produce optimal and desired tooth movement.

McGuinness et al (1992)[69] conducted a finite element analysis (FEA) to determine the stress induced in the periodontal ligament in 3 dimensions when a maxillary canine tooth is subjected to and orthodontic force similar to that produced by an edgewise appliance. The findings suggested that even with the perfect edgewise mechanics it would be difficult to obtain canine movement by pure translation or bodily movement.

Cobo et al (1993)[19] determined the stress that appears in tooth, periodontal ligament and alveolar bone, when a labiolingual force of 100 gm is applied in a labiolingual direction in a midpoint of the crown of an inferior digitalized canine, and its changes depending on the degree of loss of the supporting bone. After applying the labiolingual force in the canine, a progressive increase of the stress in the labial and lingual zones of the tooth, periodontal membrane and alveolar bone was observed when the alveolar bone was reducing. In the mesial and distal zones, no compensating forces appeared which could provoke a tooth rotation during the tipping movements.

Katona et al (1994)[56] developed a finite element model (FEM) of an orthodontic bracket bonded to enamel with GIC. The primary purpose of this project was to ascertain the effects of load misalignment on the calculated stresses within the cement layer. The results indicated that peak stress values increase as the load deflection angulation increases. If the tensile load is inadvertently applied entirely on one wing of the bracket, the stress components nearly doubled in magnitude.

Ghosh et al (1995)[37] generated finite element models for selected ceramic brackets and graphically displayed the stress distribution in the brackets when subjected to arch wire torsion and tipping forces. Six commercially available ceramic brackets, one monocrystalline and five polycrystalline alumina, of twin bracket design for the permanent maxillary left central incisor were studied. Three-dimensional computer models of the brackets were constructed and loading forces, similar to those applied by a full-size (0.0215 × 0.028 inch) stainless steel arch wire in torsion and tipping necessary to fracture ceramic brackets, were applied to the models. The brackets with an isthmus connecting the wings seemed to resist stresses better than the one bracket that did not have this feature. The design of the isthmus for the Transcend (Unitek/3M, Monrovia, Calif.) and Lumina (Ormco, Glendora, Calif.) brackets were found to be acceptable as well. The Starfire bracket ("A" Company, San Diego, Calif.) showed high stresses and irregular stress distribution, because it had sharp angles, no rounded corners, and no isthmus. The finite element method proved to be a useful tool in the stress analysis of ceramic orthodontic brackets subjected to various forces. This analysis provides key information to the development of an optimum bracket design.

Cobo et al (1996)[18] studied the stress that appears in the tooth, the periodontal ligament, and the alveolar bone, when a couple and horizontal forces were applied to obtain the bodily movement of a lower digitalized canine and its changes depending on the degree of loss of the supporting bone. The analysis of tensions was carried out by means of the finite element method (FEM) with no bone loss and after reducing the support bone 2, 4, 6, and 8 mm. When the bone loss is 2 mm, an increased stress in the levels next to the alveolar crest is already apparent. After 4, 6, and 8 mm of bone support reduction, a change of the sign and an increment of the magnitude of stress in the lowest levels occurs.

Middleton et al (1996)[73] reported an initial time-dependent (continuous/dynamic) finite element model for tooth movement that uses newly developed software, the results being cross-referenced against historical data. These early results, from a two-dimensional mathematical model of a loaded canine tooth, suggest that the remodeling process may be controlled by the periodontal ligament rather than the bone. In the finite element model, bone was found to experience a low strain of 1×10^{-5}, whereas the periodontal ligament experienced a strain of 0.1 when the "tooth model" is loaded. Only this latter figure is above the threshold usually reported to be necessary to initiate the remodeling process.

Puente et al (1996)[86] analyzed the distribution of the stress on dental and periodontal structures when a simple tipping dental movement or torque movement is produced. A tridimensional computer model based on finite element techniques was used for this purpose. The model of the lower canine was constructed on the average anatomical morphology and 396 isoparametric elements were considered. The three principal stresses (maximum, minimum and intermediate) and Von Mises stress were determined at the root, alveolar bone and periodontal ligament (PDL). It was observed how the distribution of stress is not the same for the three structures studied. In all loading cases for bucco-lingually directed forces, the three principal stresses were very similar in the PDL. The dental apex and bony alveolar crest zones are the areas that suffer the greatest stress when these kind of movements are produced.

Raboud et al (1997)[87] conducted a numerical method to provide quantitative insight into three dimensional effects for typical appliance designs. Concluded, that the out-of-plane effects are independent of the in-plane behavior so that the usual forces and moment to force ratios are maintained.

Jeon et al (1999)[52] simulated the stress response in the periodontium of the maxillary first molar to different moment to force ratios, and to determine the moment to force ratio for translational movement of the tooth by means of the finite element method. Their results demonstrated the sensitivity of the periodontium to load changes. The stress pattern in the periodontal ligament for a distalizing force without counterbalancing moments showed high concentration at the cervical level of the distobuccal root due to tipping and rotation of the tooth. After various counter rotation as well as counter tipping moments were applied, an even distribution of low compression on the distal side of the periodontal ligament was obtained at a counter tipping moment to force ratio of 9:1 and a counter rotation moment to force ratio of 5:1. Furthermore, high stress concentration was observed on the root surface at the furcation level in contrast with anterior teeth reported to display high concentration at the apex. This result may suggest that the root morphology of the maxillary first molar makes it less susceptible to apical root resorption relative to anterior teeth during tooth movement.

Thomas et al (1999)[107] reported that the tests commonly used for the evaluation of orthodontic adhesives measure tensile and shear bond strength. The two methods were compared with finite element analysis using a three-dimensional model and the effect of misalignment of the tensile and shear forces were calculated. Applying a shear load produced significant compressive and tensile stresses in the adhesive layer. Under ideal conditions of shear loading, the induced tensile stress is over 5 times the induced shear stress. The model showed that a tensile load induces predominantly tensile stresses in the adhesive layer. The calculations indicate that the tensile test method is a robust testing method with low sensitivity to misalignment of the applied load.

Geramy (2000)[33] studied the behavior of initial tooth displacements associated with alveolar bone loss situations when loaded by a force of 1 N. The results revealed that the moment/force ratio (at the bracket level) required for producing bodily movement increases in association with alveolar bone loss. Bone loss causes center of resistance movement toward the apex, but its relative distance to the alveolar crest decreases at the same time. Center of rotation of the tipping movement also shifted toward the cervical line. Among the many differences between orthodontic treatment of an adolescent and an adult patient is the presence of alveolar bone loss in the adult cases. Alveolar bone loss causes center of resistance changes as a result of the alterations in bone support. This necessitates modifications in the applied force system to produce the same movement as in a tooth with a healthy supporting structure.

Jeon et al (2001)[53] studied the use of finite element method to simulate the effect of alveolar bone loss on orthodontically induced stress in the periodontal ligament of the maxillary first molar. An anterior force of 300 g was applied at the center of the buccal crown surfaces of teeth with normal bone height and with bone loss that ranged from 2.0 to 6.0 mm. The results showed that force magnitude required lowering from 80% (2-mm bone loss) and gradually to 37% (6-mm bone loss) of the initial load applied to the tooth without bone loss. The counter tipping moment (gram-millimeters) to force (gram) ratio should increase from 9 (no bone loss) to nearly 13 (6-mm bone loss) to maintain the same range of stress in the periodontal ligament as was obtained without bone loss. A linear relationship was observed between the amount of bone loss, the desired reduction in force magnitude, and the increase in M/F ratio. The results of this study indicate that a combination of force reduction and increased M/F ratio is required to achieve uniform stress in the periodontal ligament of a tooth with bone loss.

Knox et al (2001)[58] evaluated the influence of bracket base mesh geometry on the stresses generated in the bracket-cement-tooth continuum by a shear/peel load case. When the double-mesh bracket base was considered, the combined mesh layers resulted in a decrease in the stresses recorded in the most superficial (coarse) mesh layer and an increase in the stresses recorded in the deepest (fine mesh) layer when compared with the single-layer designs in isolation. Modification of single-mesh spacing and wire diameter influences the magnitude and distribution of stresses within the bracket-cement-tooth continuum. The use of a double mesh design results in a reduction in the stresses recorded in the most superficial mesh. Mesh design influenced stress distribution in this study, primarily by determining the flexibility of the bracket base.

Rudolph et al (2001)[91] conducted a study to determine the types of orthodontic forces that cause high stress at the root apex. The material properties of enamel, dentin, PDL, and bone and 5 different load systems (tipping, intrusion, extrusion, bodily movement, and rotational force) were tested. 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; however, it was concentrated more at the alveolar crest. They conclude that intrusive, extrusive, and rotational forces produce more stress at the apex. Bodily movement and tipping forces concentrate forces at the alveolar crest, not at the apex.

Melsen (2001)[72] studied the tissue reaction to a force system generating translation of premolars and molars in the five Macaca fascicularis monkeys is described. Three force levels, 100, 200, and 300 cN were applied for a period of 11 weeks. Based on these results and a finite element model simulating the loading, a new hypothesis regarding tissue reaction to change in the stress strain distribution generated by orthodontic forces is suggested. The direct resorption could be perceived as a result of lowering of the normal strain from the functioning periodontal ligament (PDL) and as such as a start of remodelling, in the bone biological sense of the word. Indirect remodelling could be perceived as sterile inflammation attempting to remove ischaemic bone under the hyalinised tissue. At a distance from the alveolus, dense woven bone was observed as a sign of a regional acceleratory phenomena (RAP). The results of the intrusion could, according to the new hypothesis, be perceived as bending of the alveolar wall produced by the pull from Sharpey's fibres.

Schneider et al (2002)[93] studied the optimal force system for bodily movement of a single-root tooth with an orthodontic bracket attached. This was achieved by the use of the numerical finite element method, including a distinct mechanical bone-remodeling algorithm. This algorithm works with equilibrium iterations separated in 2 calculation steps. Furthermore, a parametric 3-dimensional finite element model, which allows modifications in the root length and its diameter, is described. For different geometries, the ideal moment-by-force ratios that induce a bodily movement were determined. The knowledge of root geometry is important in defining an optimal force system.

Geramy (2002)[34] investigated the stress components (S1 and S3) that appear in the periodontal membrane (PDM), when subjected to transverse and vertical loads equal to 1 N. A further aim was to quantify the alteration in stress that occurs as alveolar bone is reduced in height by 1, 2.5, 5, 6.5, and 8 mm, respectively. Six three-dimensional (3D) finite element models (FEM) of a human maxillary central incisor were designed. The models were of the same configuration except for the alveolar bone height. The results showed that alveolar bone loss caused increased stress production under the same load compared with healthy bone support (without alveolar bone resorption). Tipping movements resulted in an increased level of stress at the cervical margin of the PDM in all sampling points and at all stages of alveolar bone loss. These increased stress components were found to be at the sub-apical and apical levels for intrusive movement.

Kang et al (2003)[55] analyzed the relationship between the critical contact angle and the torque angle in an orthodontic bracket and archwire assembly in 3 dimensions. Three-dimensional mathematical models were created with geometric bracket-archwire parameters that included 2 slot sizes, 3 bracket widths, and 3 to 4 wire sizes. From this, 3-dimensional mathematical equations (3DMEs) for the critical contact angle and the maximum torque that result in critical contact angles of 0º were derived and calculated. For all bracket-archwire combinations, the critical contact angle decreased as bracket width, torque angle, and wire size increased. Therefore, all bracket-archwire parameters except slot height had an effect on the critical contact angle. In addition, the effect of a beveled edge was investigated in some archwires. The results of this study provide theoretic and experimental bases for clinical orthodontic practice and indicate that torque angles should be included in the evaluation of the critical contact angle.

Toms et al (2003)[108] sought to determine the importance of using nonlinear mechanical properties and non-uniform geometric data in computer predictions of periodontal ligament stresses and tooth movements. Predictions of the maximum and minimum principal stresses and von Mises stresses in the PDL were determined for extrusive and tipping forces. The results indicated that biofidelic finite element models predicted substantially different stresses in the PDL for extrusive loading than did the uniform thickness model, suggesting that incorporation of the hourglass shape of the PDL is warranted. In addition, incorporation of nonlinear mechanical properties for the PDL resulted in dramatic increases in the stresses at the apex and cervical margin as compared with the linear models.

Cattaneo et al (2003)[14] conducted a FEA that allowed them to simulate the displacement of a molar in relation to the well-defined morphology of the maxilla. When the molar was loaded with occlusal forces, the stresses were transferred predominantly through the infrazygomatic crest. This changed when mesial and distal displacements of the molars were simulated. In the model with mesial molar displacement, a larger part of the bite forces were transferred through the anterior part of the maxilla, resulting in the buccal bone being loaded in compression. In the model with distal molar displacement, the posterior part of the maxilla was deformed through compression; this resulted in higher compensatory tensile stresses in the anterior part of the maxilla and at the zygomatic arch. This distribution of the occlusal forces might contribute to the posterior rotation often described as the orthopedic effect of extraoral traction.

Kojimaa et al (2005)[59] discussed a method that allowed the simulation of more complex tooth movements. A 3-dimensional finite element method was used to simulate the orthodontic tooth movement (retraction) of a maxillary canine by sliding mechanics and any associated movement of the anchor teeth. Absorption and apposition of the alveolar bone were produced in proportion to the stress of the periodontal ligament. The canine tipped during the initial unsteady state and then moved bodily during the steady state. It became upright when the orthodontic force was removed. The anchor teeth moved in the steady state and tipped in the mesial direction. The decrease in applied force by friction was about 70%. The tipping of the canine decreased when the wire size was increased or when the applied force was decreased. They suggested that this method might enable one to estimate various tooth movements clinically.

Ziegler et al (2005)[120] studied the elastic properties of the periodontal ligament (PDL) in eight multi-rooted teeth were examined in a combined experimental and numerical study in six minipigs. The initial tooth movement of the mandibular primary molars surrounded by the periodontium was registered three-dimensionally (3D) in an optomechanical measuring system. The dissections were then embedded in resin and cut in transverse sections. Based on these sections, 3D finite element (FE) models were constructed and numerically loaded with the same force systems as used in the experiment. There was no significant difference in the material parameters determined for specimens with two, four or six roots. The results were in close agreement with the material parameters of the PDL, determined in previous investigations of single-rooted human and pig teeth.

Kojimaa et al (2006)[60] developed a comprehensive mechanical, 3-dimensional, numerical model for predicting tooth movement. Tooth movements produced by wire bending were simulated numerically. The teeth moved as a result of bone remodeling, which occurs in proportion to stress in the periodontal ligament. With an off-center bend, a tooth near the bending position was subjected to a large moment and tipped more noticeably than the other teeth. Also, a tooth far from the bending position moved slightly in the mesial or the distal direction. With the center V-bend, when the second molar was added as an anchor tooth, the tipping angle and the intrusion of the canine increased, and movement of the first molar was prevented. When a wire with an inverse curve of spee was placed in the mandibular arch, the calculated tendency of vertical tooth movements was the same as the measured result. In these tooth movements, the initial force system changed as the teeth moved. Tooth movement was influenced by the size of the root surface area. Concluded, that tooth movements produced by wire bending could be estimated.

Kojima et al (2006)[61] studied the combined effect of friction and an archwire's flexural rigidity on canine movement in sliding mechanics, and to explain how to select a suitable archwire and force level for efficient bodily movement. As the frictional force decreased, both the net force acting on and the moving speed of the canine increased. The elastic deformation of the archwire increased, and the moving pattern of the canine changed from bodily movement to tipping, although there was no clearance between the archwire and the bracket slot. When a light wire was used, wire deformation increased, and the canine experienced greater tipping.

Jayade et al (2007)[51] evaluated the magnitudes of initial and subsequent sequential deactivational third order moments generated in rectangular twisted archwires in order to judge their biologic acceptability. A finite element study was carried out with the MSC Patran/Nastran interface. Required twists were applied at the appropriate locations to derive the applied and reactionary moments both initially and during the time needed for complete deactivation. The results indicated that a round-tripping possibility does exist in certain clinical procedures. Furthermore, the moments produced could be quite high, thereby enhancing the possibility of root resorption. They concluded twists in rectangular archwires may be used only when reciprocal torque is needed on adjacent teeth. In other situations, alternative torquing methods should be considered.

Hohmanna et al (2007)[45] evaluated the risk of root resorption, individual finite element models (FEMs) of extracted human maxillary first premolars were created, and the distribution of the hydrostatic pressure in the periodontal ligament (PDL) of these models was simulated. The results of clinical examination and simulations were compared using the identical roots of the teeth. The regions that showed increased hydrostatic pressure correlated well with the locations of root resorption for each tooth. Increased torque resulted in increased high-pressure areas and increased magnitudes of hydrostatic pressure, correlating with the experiments. Thus, concluded if hydrostatic pressure exceeds typical human capillary blood pressure in the PDL, the risk of root resorption increases.

Reimann et al (2007)[90] investigated the combined Centre of Resistance (CR) of the upper four incisors numerically using finite-element (FE) method. In the FE system, the model of the anterior segment was loaded with torques of 10 Nmm each at the lateral incisors. The FE model indicated that the individual incisors moved independently, although they were blocked with a steel wire of dimension $0.46 \times 0.65\ mm^2$. The individual CRs were located at 5 mm distal and 9 and 12 mm apical to the centre of the lateral brackets. Thus, the classical view of a combined CR for the anterior segment was disproved and the planning of orthodontic tooth movements of the upper incisors should no longer be based on that concept.

Ulusoya et al (2008)[110] evaluated the effects of the Class II activator and the Class II activator high-pull headgear (HG) combination on the mandible with 3-dimensional (3D) finite element stress analysis. To investigate the effects of the Class II activator, a 3D model of the lower part of this appliance was constructed and fixed on the mandibular model. The Class II activator high-pull headgear model was established as described, and an extraoral traction force of 350 g was directed from the middle of the Class II activator to the top of the mandibular condyle. The stress regions were studied with the finite element method. The regions near the muscle attachment areas were affected the most. The inner part of the coronoid process and the gonial area had the maximum stress values.

Therefore, both functional appliances can cause morphologic changes on the mandible by activating the masticatory muscles to change the growth direction.

Cattaneo et al (2008)[15] demonstrated by FE analyses that the influence of the material properties of the PDL on the type of tooth movement. Moreover, the influence of the applied force level on the type of tooth movement, with a fixed M/F ratio, was evaluated and the results interpreted in the light of existing prescriptions for orthodontic tooth movement. By applying a range of values of M/F, different types of tooth movement were generated, although the classic prescription of the M/F ratio suggested in the literature could not be confirmed. Due to the nonlinear behavior of the periodontal ligament, loading modes with a constant M/F ratio, yet varying the force magnitude, resulted in different types of tooth movement. Therefore**,** the material properties of the periodontal ligament, the morphology of the root, and the alveolar bone are patient specific. Therefore, the M/F values generally advocated to obtain orthodontic tooth movement should be used only as guidelines. To be effective and accurate, the force system selected for a specific tooth movement must be monitored and the outcome compared with the predicted tooth movement.

Holberg et al (2008)[46] analyzed the strains induced in the sutures of the midface and the cranial base by headgear therapy involving orthopedic forces. A finite element model of the viscerocranium and the neurocranium was used. The magnitude and the distribution of the measured strains depended on the level and the direction of the acting force. Overall, the strain values measured at the sutures of the midface and the cranial base were moderate. The measured peak values at a load of 5 N per side were usually just below 20µ strain irrespective of the force direction. A characteristic distribution of strain values appeared on the anatomical structures of the midface and the cranial base for each vector direction. The measurements based on the finite element method provided a good overview of the approximate magnitudes of sutural strains with orthopedic headgear therapy. The signal arriving in the sutures is apparently well below threshold, since the maximum measured strains in most sutures were about 100 fold lower than the minimal effective strain. A skeletal effect of the orthopedic headgear due to a mechanical effect on sutural growth cannot be confirmed from these results. They concluded that the good clinical efficacy of headgear therapy with orthopedic forces is apparently based mainly on dentoalveolar effects, whereas the skeletal effect due to inhibition of sutural growth is somewhat questionable.

Provatidis et al (2008)[85] did a finite element model (FEM) of a dry human skull with the RME appliance cemented in place in order to evaluate these effects on the overall craniofacial complex with different suture ossification. The behaviour of the FEM was compared with the findings of a clinical study and to an in vitro experiment of the same dry skull. It was found that the maxillolacrymal, the frontomaxillary, the nasomaxillary, the transverse midpalatal sutures, and the suture between the maxilla and pterygoid process of the sphenoid bone did not influence the outcome of RME, while the zygomatico-maxillary suture influenced the response of the craniofacial complex to the expansion forces. Moreover, the sagittal suture at the level of the frontal part of the midpalatal suture plays an important role in the degree and manner of maxillary separation.

Gautam et al (2009)[32] evaluated biomechanically the displacement patterns of the facial bones in response to different headgear loading by using a higher-resolution finite element method model than used in previous studies. Different headgear forces were simulated by applying 1 kg of posteriorly directed force in the first molar region to simulate cervical-pull, straight-pull, and high-pull headgear. The distal displacement of the maxilla was the greatest with the straight-pull headgear followed by the cervical-pull headgear. The high-pull headgear had better control in the vertical dimensions. The center of rotation varied with the direction of headgear forces for both the maxilla and the zygomatic complex. A potential for chondrogenic and osteogenic modeling exists for the articular fossa and the articular eminence with headgear loading.

Wei et al (2009)[115] conducted a study to provide the lingual technique with valuable information by using a 3-dimensional (3D) finite element method (FEM). Horizontal retraction force, vertical intrusive force, and lingual root torque were applied to simulate labial and lingual orthodontic treatment. Loads of the same magnitude produced translation of the maxillary incisor in labial orthodontics but lingual crown tipping of the same tooth in lingual orthodontics. This suggests that loss of torque control of the maxillary incisors during retraction in extraction patients is more likely in lingual orthodontic treatment. Therefore, Lingual orthodontics should not simply follow the clinical experience of the labial techniques but should increase lingual root torque, increase vertical intrusive force, and decrease horizontal retraction force properly to achieve the best orthodontic results.

Huang et al (2009)[47] investigated the torque capabilities of conventional and self-ligating brackets by using the finite element method. Three types of brackets were selected: self-ligating Hanson Speed and Damon MX, and conventionally ligated Discovery. Torque of 20° was applied to the maxillary right incisor with 0.46 X 0.64mm2 (0.018 X 0.025 in) and 0.48 X 0.64 mm2 (0.019 X 0.025 in) archwires. Three kinds of wire alloys were used: stainless steel, titanium molybdenum, and nickel titanium. For the conventional Discovery brackets, 2 types of ligation were modeled: elastic and stainless steel wire ligatures. The torque angle/torque moment curves seemed to be dominated by the characteristics of the wire. The change of wire dimension increased the torque moments less than the change of wire alloy (125% increase for a 0.48X0.64mm2 instead of a 0.46X0.64mm2 stainless steel wire, and 220% for a 0.46 X 0.64 mm2 stainless steel instead of a nickel-titanium wire). The combined change of the wire alloy and wire dimension resulted in a 600% increase for a 0.48 X 0.64 mm2 stainless steel instead of a 0.46 X 0.64 mm2 nickel-titanium wire. The play of the 0.46 X 0.64 mm2 wires was about 9.0°, and the play of the 0.48X0.64mm2 wires was about 7.5°, with slightly more play for the Damon. Therefore, improving the adaptation of torque movements to the biomechanical reactions of the periodontium is best done by proper selection of both wire dimension and wire alloy. The effect of the bracket system is of minor importance, with the exception of brackets with an active clip (eg, Speed), which had the least play and the lowest torquing moments of all the wires.

Kojimaa et al (2010)[61] calculated the long-term tooth movements in en-masse sliding mechanics. Long-term tooth movements in en-masse sliding mechanics were simulated with the finite element method. Tipping of the anterior teeth occurred immediately after application of retraction forces. The force system then changed so that the teeth moved almost bodily, and friction occurred at the bracket-wire interface. Irrespective of the amount of friction, the ratio of movement distances between the posterior and anterior teeth was almost the same. By increasing the applied force or decreasing the frictional coefficient, the teeth moved rapidly, but the tipping angle of the anterior teeth increased because of the elastic deflection of the archwire. Finite element simulation clarified the tooth movement and the force system in en-masse sliding mechanics.

Xua et al (2011)[116] determined the elastic modulus of the periodontal ligament (PDL). The study was carried out on eight human maxillary jaw segments containing central incisors. Displacements were measured under load using a electronic speckle pattern interferometry (ESPI). Subsequently, FEM presenting the same individual geometry as the respective autopsy material were developed to simulate tooth mobility numerically under the same force systems as were used in the experiment. A bilinear material parameter set was assumed to simulate tooth deflections. Thus, the force/deflection curves from the measurements showed a significant nonlinear behavior of elastic stiffness of the PDL.

2) TORQUE AND SELF LIGATING BRACKETS IN ORTHODONTICS

Rauch (1959)[88] stated that in order to attain our present-day goals of treatment, a definite technique for the application of torque force becomes imperative. The orthodontist will experience little difficulty if he will keep in mind the following fundamental principles: the crown of a tooth moves in the direction of torque; the root of a tooth moves in the opposite direction of torque; and, by the application of an auxiliary force derived from elastics or other sources, this torque action can he altered in such a way as to cause either the root or the crown of a tooth to move in whichever direction the operator may desire.

Germane et al (1989)[35] studied the facial surface contours of 600 maxillary and mandibular teeth, including 50 of each type of tooth from central incisors to first molars, were measured. The magnitude of the variation found was so great as to suggest that differences between patients or differences in height of bracket placement are greater than the differences between the standard torque prescriptions now used in orthodontics. No single point, including the coronal midpoint (LA point), was found to be constant among teeth of the same type. Variation in facial surface contour tended to be greater in the posterior teeth than in the anterior teeth. Future custom construction of brackets, adjusted to individual facial contour differences, will also require information regarding optimal tooth position in the head, including compensations necessary for variations in facial skeletal pattern.

Creekmore et al (1993)[20] stated that the frequently anticipated results of treatment are not achieved by using preadjusted appliances and straight wires. This is due to inaccurate bracket placement, variations in tooth structure, variations in the maxillary/mandibular relationships, tissue rebound, and mechanical deficiencies of edgewise orthodontic appliances. Beyond the accuracy or inaccuracy of bracket placement and the fact that brackets are placed away from the center of resistance, orthodontic appliances have two additional significant mechanical deficiencies; play between the arch wire and the arch wire slot, and force diminution. These deficiencies cannot be eliminated from current appliances, however, they can be minimized by using reasonably stiff arch wires approximating the size of the arch wire slots. The amount of play plus the amount of force diminution inherent in your appliance can be added to or subtracted from the torque, tip, rotation, and height parameters for each bracket to deliver the teeth to the desired positions. Therefore treatment goals can be achieved with maximum efficiency.

Isaacson et al (1993)[48] reported traditional edgewise orthodontic mechanics are significantly limited in their ability to provide incisor torque control because of the limitations of bracket-to-bracket mechanics and the poorly defined reciprocal actions inherently produced. The science of mechanics dictates that all incisor torque control mechanisms must act through one of two basic principles: the moment of a couple or the moment of a force. The torquing arch is a modification of the traditional edgewise system and employs the moment of a couple to achieve incisor torque control and precise definition of reciprocal effects. Alternatively, the base arch uses the moment of a force to also rotate incisors in a crown facial/root lingual direction. The base arch, however, includes a large moment to rotate molars in a crown distal/root mesial direction, and concurrent equilibrium forces to intrude incisors and extrude molars. Depending on how they are employed, torquing arches and base arches may also rotate molars in a faciolingual direction, enhance or diminish posterior anchorage, and increase or conserve arch perimenter.

Odegaard et al (1994)[79] demonstrated that the amount of play between bracket and wire in torsion for individual tooth movement is considerably larger than the amount expected. It has also been shown that the initial portion of the load/deflection curves are relatively flat for the smaller dimensions before a linear relationship between moment and deflection is achieved, indicating a restraining effect caused by the ligature. The resulting curves using wire "a" without ligature illustrates this point. The linear portions of the curves show that the change in effective rotational moment will change rapidly for small changes in the tooth axial inclination, suggesting that reactivation of the wires should take place at frequent intervals. For individual tooth torque, a more efficient method can be the use of highly elastic wires in combination with brackets with variable torque.

Shivapuja et al (1994)[95] reported the increased use of self-ligating bracket systems frequently raises the question of how they compare with conventional ligation systems. An in vitro and clinical investigation was undertaken to evaluate and compare these distinctly different groups, by using five different brackets. The Activa ("A" Company, Johnson & Johnson, San Diego, Calif.), Edgelok

(Ormco, GIendora, Calif.), and SPEED (Strite Industries Ltd., Cambridge, Ontario) self-ligating bracket systems displayed a significantly lower level of frictional resistance, dramatically less chairtime for arch wire removal and insertion, and promoted improved infection control, when compared with polyurethane elastomeric and stainless steel tie wire ligation for ceramic and metal twin brackets.

Harradine (2003)[42] reported that currently available self-ligating brackets offer the very valuable combination of extremely low friction and secure full bracket engagement and, at last, they deliver most of the potential advantages of this type of bracket. These developments offer the possibility of a significant reduction in average treatment times and also in anchorage requirements, particularly in cases requiring large tooth movements. Whilst further refinements are desirable and further studies essential, current brackets are able to deliver measurable benefit with good robustness and ease of use.

Harzer (2004)[43] investigated slot deformation and the equivalent torque capacity of polycarbonate brackets with and without a metal slot in comparison with those of a metal bracket. For this purpose, the expansion characteristics and, in a further investigation, the labial crown torque of an upper central incisor, were measured in a simulated intra-oral clinical situation, using the orthodontic measuring and simulation system (OMSS). Three types of bracket with a 0.018 inch slot were tested: polycarbonate Brillant without a metal slot, Elegance with a metal slot and the metal bracket, Mini-Mono. For testing purposes the brackets were torqued with 0.016 × 0.022 inch and 0.018 × 0.022 inch ideal stainless steel archwires. In the activating experiments, significantly higher torque losses and lower torquing moments were registered with both rectangular archwires with the polycarbonate brackets than with the metal bracket. In the simulation tests, significantly higher torquing moments were registered with the metal bracket than with the polycarbonate brackets. On the basis of the present results, all three brackets can be recommended for torquing. However, in view of the high torque losses, the torques programmed in the straightwire technique must be seen as questionable.

Cash et al (2004)[13] evaluated the slots of five upper left central incisor brackets from 11 commercially available bracket systems of 0.022-inch (0.5588 mm) dimension. Results indicate that all bracket slots are oversized. Three bracket systems slots (Twin Torque, Clarity, and Mini Mono) were within 5% (61.08, 1.655, 1.75) of their stated dimensions with essentially parallel slot walls. The Elegance Plastic slot was parallel sided but oversized by 12% (61.15). The geometry of bracket slots was also variable. The Victory Series slot was slightly divergent with the top oversized by 6% (61.035). The Nu-Edge slot was divergent and slot top oversized by 14% (61.32). The Mxi Advant-Edge, Damon II SL, Elite Mini Opti-MIM Roth, and MBT were all convergent, and the base of the Damon slot was oversized by 17% (61.79). The Discovery bracket was convergent, and the slot base was oversized by 24% (61.255), which was the largest recorded variance. This bracket also had a 7% difference between the widths of the slot top and the base. Inaccurate machining of bracket slot dimensions and the use of undersized archwires may directly and adversely affect three-dimensional tooth positioning.

Pandis et al (2006)[83] A randomized clinical trial done that the engagement mode of wire to bracket affects the buccolingual inclination of maxillary incisors in extraction and non-extraction treatment with self-ligating and conventional brackets. Difference in the buccolingual inclination of maxillary incisors before and after treatment with the two appliances across the two treatment groups (extraction and non-extraction). Angular measurements of the Sella-Nasion and Nasion-A point to maxillary incisor axis was calculated. No difference was found in the mean difference of the two angles measured for the two bracket groups studied. Self-ligating brackets seem to be equally efficient in delivering torque to maxillary incisors relative to conventional brackets in extraction and non-extraction cases.

Pandis et al (2007)[81] investigated the effect of intraoral aging on the force applied during engagement of a wire into the slot of active self-ligating brackets. Two types of brackets were used: Speed and In Ovation-R. No difference was found between as-received and used brackets with respect to force exerted by the spring in 1 bracket group, whereas the other group showed extensive relaxation after use; neither group had permanent deformation. The consistency of the initial force levels varied significantly in each bracket group. Thus, the initial force levels and the effect of intraoral conditions on the stiffness of the clip seem to vary between products, with potential implications for the archwire engagement into the bracket slot and associated mechanotherapy.

Turnbull et al (2007)[109] conducted a prospective clinical study, where they assessed the relative speed of archwire changes, comparing self-ligating brackets with conventional elastomeric ligation methods, and further assessed this in relation to the stage of orthodontic treatment represented by different wire sizes and types. The main outcome measure was the time to remove or place elastomeric ligatures or open/close self-ligating brackets for 2 matched groups of fixed appliance patients: Damon2 self-ligating bracket (SDS Ormco, Orange, Calif) and a conventional mini-twin bracket (Orthos, SDS Ormco). The Damon2 self-ligating system had a significantly shorter mean archwire ligation time for both placing and removing wires compared with the conventional elastomeric system. Ligation of an archwire was approximately twice as quick with the self-ligating system. The type of bracket and the size of wire used are statistically significant predictors for speed of ligation and chairside time. The self-ligating system offered quicker and arguably more efficient wire removal and placement for most orthodontic treatment stages.

Streva et al (2007)[99] verified the torque precision of metallic brackets with MBT prescription using the canine brackets as the representative sample of six commercial brands. Twenty maxillary and mandibular canine brackets of one of the following commercial brands were selected: 3M Unitek, Abzil, American Orthodontics, TP Orthodontics, Morelli and Ortho Organizers. The results showed that for the maxillary canine brackets, only the Morelli torque (-3.33º) presented statistically significant difference from the proposed values (-7º). For the mandibular canines, American Orthodontics (-6.34º) and Ortho Organizers (-

6.25º) presented statistically significant differences from the standards (-6º). Comparing the brands, Morelli presented statistically significant differences in comparison with all the other brands for maxillary canine brackets. For the mandibular canine brackets, there was no statistically significant difference between the brands. There are significant variations in torque values of some of the brackets assessed, which would clinically compromise the buccolingual positioning of the tooth at the end of orthodontic treatment.

Morina et al (2008)[76] investigated the torque capacity of active and passive self ligating brackets compared with metallic, ceramic, and polycarbonate edgewise brackets. Six types of orthodontic brackets were included in the study: the self-ligating Speed and Damon2, the stainless steel (SS), Ultratrimm and Discovery, the ceramic bracket, Fascination 2, and the polycarbonate bracket, Brillant. All brackets had a 0.022-inch slot size and were torqued with 0.019 × 0.025-inch SS archwires. For this purpose, the labial crown torque of an upper central incisor was measured in a simulated intraoral clinical situation using the orthodontic measurement and simulation system (OMSS). A torque of 20 degrees was applied and the correction of the misalignement was simulated experimentally with the OMSS. The ceramic bracket (Fascination 2) presented the highest torquing moment (35 Nmm) and, together with a SS bracket, the lowest torque loss (4.6 degrees). Self-ligating, polycarbonate, and selective metallic brackets demonstrated almost a 7-fold decreased moment developed during insertion of a 0.019 × 0.022- inch SS wire into a 0.022-inch slot and a 100 per cent increase in loss.

Badawi et al (2008)[6] measured the difference in third-order moments that can be delivered by engaging 0.019 X 0.025-in stainless steel archwires to 2 active self-ligating brackets (In-Ovation, GAC, Bohemia, NY; Speed, Strite Industries, Cambridge, Ontario, Canada) and 2 passive self-ligating brackets (Damon2, Ormco, Orange, Calif; Smart Clip, 3M Unitek, Monrovia, Calif). A bracket/wire assembly torsion device was developed. There was a significant difference in the engagement angle between the 2 types of brackets; on average, torque started to be expressed at 7.5° of torsion for the active self-ligating brackets and at 15° of torsion for the passive self-ligating brackets. The torque expression was higher for the active self-ligating brackets up to 35° of torsion. Torsion of the wire past this point resulted in a linear increase of the measured torque for the Damon2, the Smart Clip, and the In-Ovation brackets. The torque was relatively constant past 35° of torsion for the Speed bracket. They concluded that active self-ligating brackets are more effective in torque expression than passive self-ligating brackets.

Nishio et al (2009)[77] evaluated the resistance to deformation or fracture of esthetic brackets produced by archwire torsion. Six types of maxillary right central incisor brackets were analyzed: traditional ceramic brackets (cer); ceramic brackets reinforced with a stainless steel slot (cer/ss); ceramic brackets reinforced with a gold slot (cer/gold); traditional polycarbonate brackets (poly); polycarbonate brackets reinforced with a stainless steel slot (poly/ss); and polycarbonate brackets reinforced with ceramic fillers and a stainless steel slot (poly/cer/ss). They suggested that the stainless steel slot might enhance resistance to deformation or fracture, although gold slots and ceramic fillers are ineffective for reinforcing esthetic brackets.

Pandis(2009)[82] comparatively assessed the magnitude and direction of forces and moments generated from different bracket systems, during the initial levelling and alignment stage of orthodontic treatment. Three types of brackets were used: Orthos2 (Ormco), Damon2 (Ormco), and In-Ovation R (GAC). The model was mounted on the Orthodontic Measurement and Simulation System (OMSS) and six static measurements were taken at the initial crowded state per bracket for the lateral incisor, canine, and first premolar. The lingually inclined, crowded lateral incisor presented an extrusive and buccal movement and showed the lowest force in the vertical direction, whereas the self-ligating group of brackets generated the highest force in the buccolingual direction. The moments applied by the three bracket systems followed the general trend shown for forces; in the vertical axis, the self-ligating brackets exerted lower forces than their conventional counterpart. This was modified in the buccolingual direction where, in most instances, the self-ligating appliances applied higher moments compared with the conventional bracket. In most cases, the magnitude of forces and moments ranged between 30 – 70 cN and 2 – 6 N mm, respectively. However, maximum forces and moments developed at the lateral incisor were almost four times higher than the average.

Chung et al (2009)[17] examined the influence of third-order torque on kinetic friction in sliding mechanics involving active and passive self-ligating brackets. Wire-slot frictional forces were quantified and compared across five sets of brackets and tubes within a simulated posterior dental segment with -15, -10, -5, 0, +5, +10, and +15 of torque placed in the second-premolar bracket; a working archwire was pulled through the slots. They concluded that third-order torque in posterior dental segments can generate frictional resistance during anterior retraction with the archwire sliding through self-ligating bracket slots. With small torque angles, friction is less with passive than with active self-ligating brackets, but bracket design is a factor. Frictional forces are substantial, regardless of ligation if the wire-slot torque exceeds the third-order clearance.

Pandis et al (2010)[80] compared the time required to complete the alignment of crowded maxillary anterior teeth (canine to canine) between Damon MX (Ormco, Glendora, Calif) and In-Ovation R (GAC, Central Islip, NY) self-ligating brackets. No difference in crowding alleviation was found between the 2 bracket systems. Higher irregularity index values were associated with the increased probability of delayed resolving of crowding. Conclusions: The use of passive or active self-ligating brackets does not seem to affect treatment duration for alleviating initial crowding.

Chen et al (2010)[16] conducted a systematic review to identify and review the orthodontic literature with regard to the efficiency, effectiveness, and stability of treatment with self ligating brackets compared with conventional brackets. Sixteen studies met the inclusion criteria, including 2 randomized controlled trials with low risk of bias, 10 cohort studies with moderate risk of bias, and 4 cross sectional studies with moderate to high risk of bias. Self-ligation appears to have a significant advantage with regard to chair time, based on several cross-sectional studies. Analyses also showed a small, but statistically significant, difference in mandibular incisor proclination. No other differences in treatment time and occlusal characteristics after treatment were found between the 2 systems. No studies on long-term stability of treatment were identified. They concluded that despite claims about the advantages of self-ligating brackets, evidence is generally lacking. Shortened chair time and slightly less incisor proclination appear to be the only significant advantages of self-ligating systems over conventional systems that are supported by the current evidence.

Archambault et al (2010)[4] evaluated the quantitative effects on torque expression of varying the slot size of stainless steel orthodontic brackets and the dimension of stainless steel wire, and to analyze the limitations of the experimental methods used. In vitro studies measuring torque expression in conventional and self ligating stainless steel brackets with a torque-measuring device, with the use of straight stainless steel orthodontic wire without second-order mechanics and without loops, coils, or auxiliary wires, were sought through a systematic review process. On the basis of the selected studies, in a 0.018 inch stainless steel bracket slot, the engagement angle ranges from 31 degrees with a 0.016 X 0.016 inch stainless steel archwire to 4.6 degrees with a 0.018 X 0.025 inch stainless steel archwire. In a 0.022 inch stainless steel bracket slot, the engagement angle ranges from 18 degrees with a 0.018 X 0.025 inch stainless steel archwire to 6 degrees with a 0.021 X 0.025 inch stainless steel archwire. Active stainless steel self-ligating brackets demonstrate an engagement angle of approximately 7.5 degrees, whereas passive stainless steel self-ligating brackets show an engagement angle of approximately 14 degrees with 0.019 X 0.025 inch stainless steel wire in a 0.022 inch slot. They concluded that the engagement angle depends on archwire dimension and edge shape, as well as on bracket slot dimension, and is variable and larger than published theoretical values. Clinically effective torque can be achieved in a 0.022 inch bracket slot with archwire torsion of 15 to 31 degrees for active self-ligating brackets and of 23 to 35 degrees for passive self-ligating brackets with a 0.019 X 0.025 inch stainless steel wire.

Fleming et al (2010)[27] evaluated the clinical differences in relation to the use of self-ligating brackets in orthodontics. Randomized controlled trials (RCTs) and controlled clinical trials (CCTs) investigating the influence of bracket type on alignment efficiency, subjective pain experience, bond failure rate, arch dimensional changes, rate of orthodontic space closure, periodontal outcomes, and root resorption were selected. Concluded at this stage there is insufficient high-quality evidence to support the use of self ligating fixed orthodontic appliances over conventional appliance systems or vice versa.

Major et al (2010)[64] stated that in all manufacturing processes there are tolerances; however, orthodontic bracket manufacturers seldom state the slot dimensional tolerances. Their experiment developed a novel method of analyzing slot profile dimensions using photographs of the slot. Five points are selected along each wall, and lines are fitted to define a trapezoidal slot shape. This investigation measures slot height at the slot’s top and bottom, angles between walls, slot taper, and the linearity of each wall. Slot dimensions for 30 upper right central incisor self-ligating stainless steel brackets from three manufacturers were evaluated. Speed brackets have a slot height 2% smaller than the nominal 0.559mm size and have a slightly convergent taper. In-Ovation brackets have a divergent taper at an average angle of 1.47 degrees. In-Ovation is closest to the nominal value of slot height at the slot base and has the smallest manufacturing tolerances. Damon Q brackets are the most rectangular in shape, with nearly 90-degree corners between the slot bottom and walls. Damon slot height is on average 3% oversized.

Major et al (2011)[65] investigated the third-order torque on different types of self-ligated brackets by analyzing the bracket's elastic and plastic deformations in conjunction with the expressed torque at varying angles of twist. An orthodontic bracket was mounted to a load cell that measured forces and moments in all directions. The wire was twisted in the bracket via a stepper motor, controlled by custom software. At the maximum torquing angle of 63° with 0.019 X 0.025-in stainless steel wire, the total elastic and plastic deformation values were 0.063, 0.033, and 0.137 mm for Damon Q (Ormco, Orange, Calif), In-Ovation R (GAC, Bohemia, NY), and Speed (Strite Industries, Cambridge, Ontario, Canada), respectively. The total plastic deformation values were 0.015, 0.006, and 0.086 mm, respectively, measured at 0_ of unloading. Conclusions: In-Ovation R had the least deformation due to torquing of the 3 investigated bracket types. Damon Q and Speed on average had approximately 2.5 and 14 times greater maximum plastic deformation, respectively, than did In-Ovation R.

Major et al (2011)[66] conducted a study was to quantify torque expression in 3 self-ligation bracket systems (Damon Q, In-Ovation R, and Speed) during loading and unloading. A stepper motor was used to rotate a wire in a fixed bracket slot from –15º to 63º in 3º increments, and then back to –15º. The bracket was mounted on top of a load cell that measured forces and moments in all directions. Results showed that Damon's and In-Ovation's maximum average torque values at 63º were 105 and 113 Nmm, respectively. Many Speed brackets experienced premature loss of torque between 48º and 63º, and the average maximum was 82 Nmm at 54º. The torque plays for Damon, In-Ovation, and Speed were 11.3º, 11.9º, and 10.8º, respectively. Generally, In-Ovation expressed the most torque at a given angle of twist, followed by Damon and then Speed. However, there was no significant difference between brackets below 34 Nmm of torque. From a clinical perspective, the torque plays between brackets were virtually indistinguishable.

MATERIALS AND METHODS

Materials used in this study

Brackets used - Two Active and two Passive self-ligating bracket systems were selected and one conventional bracket system served as control.

1. **Active self-ligating bracket systems -**

a) In-Ovation R (GAC-Dentsply-USA)

b) Time^{TM} (American Orthodontics-USA)

2. **Passive self-ligating bracket systems -**

a) Smart Clip SL3-(3 M Unitek-USA)

b) Damon 3MX- (Ormco Orthodontics-California)

3. **Conventional ligation system -**

Ovation-(GAC-Dentsply-USA)

Upper Right Central Incisor Stainless Steel Roth Prescription bracket with slot dimension of 0.022 x 0.028 inches was used in all the 3 groups. **(Figure 1)**

Archwires used -

a) 0.017 x 0.025 inches-straight length Stainless steel wires (GAC-USA)

b) 0.019 x0.025 inches straight length Stainless steel wires (GAC-USA)

c) 0.021 x 0.025 inches- straight length Stainless steel wires (GAC-USA)

Study methodology

The steps involved in finite element analysis are:-

1) **Pre-Processing Phase**: the steps involved in this phase included -

   a) Creating a 3-dimensional model which was achieved from a computed tomography scan [**Figure 2(a)]**, and a white light scanner (**Figure 3)**.

   b) Material properties (Youngs modulus and Poissons ratio) were assigned to the elements to determine the way they will behave when the load is applied **(Table 1).**

2) **Processing/Solution Phase**: The boundary conditions in the finite element model were defined. These determine the degree of freedom of movement that is allowed for the model.

3) **Post-Processing Phase**: This is the last step which displays the results obtained from the processing/solution phase. Results were obtained in graphical, numerical or animated format.

1) PRE-PROCESSING STAGE:-

(A) 3 Dimensional Modeling of Tooth and its Supporting Structures:-

Computerized Tomography (CT) (GE Healthcare Technologies - Lightspeed VCT, Bharat Scans, Chennai.) image acquisitions in DICOM (Digital Imaging Communications in Medicine) format of an adult dry human skull was performed using 120 kV, 150 mA, 512 x

512 matrix, field of view 14 x 14 cm and slice thickness of 0.5 mm **[Figure 2(a)]**.[32,49,50]

These CT images consisted of 165 sections along the axial axis and 123 sections along the coronal axis, were imported into the software program "Pro/Engineer Wildfire Version 4.0" (Parametric Technology Corporation) **[Figure 2(b)].**

From this point, segmentation was started. **Segmentation** is a process that consists of separating the right maxillary central incisor and its supporting structures from other adjacent anatomical structures in different groups or masks, such as enamel, dentine, pulp, cortical and cancellous bone according to their radiodensities expressed in Hounsfield units **[Figure 2(c)]**.

A 3 dimensional model of the right maxillary central incisor and its supporting structures was generated by a **Computer Aided Designing/Computer Aided Engineering** (CAD/CAE) program. The creation of the periodontal ligament with a thickness of 0.25 mm was performed due to the impossibility to define this structure from CT images (pixel size = 0.273 mm). The isocurves of the cortical and cancellous bones were also exported, with their thickness being defined as 10 mm, and whose extremities corresponded to the position of the right maxillary central incisor. After the generation of a surface mesh for every structure, a volumetric mesh with tetrahedral elements was generated.[54,91,102,112]

From the curves, **Surfaces** were created using a command called **Boundaries**. From these surfaces, a **Solid** is generated. Once the tooth was developed, in a similar fashion all other parts of the periodontium were created and assembled. This assembly was then exported to the analysis package software (Ansys Workbench Version 11). The export was performed through a bidirectionally understandable translator called **Initial Graphics Exchange Specifications** (IGES). This file format of export is understandable by most of the software programs **[Figure 2(c)]**.

(B) 3 Dimensional Modeling of Brackets: - Reverse engineering is the process used for discovering the technological principles of the small components and analyzing their structure, function, and operations. Reverse engineering of the selected brackets was done by a **Comet 5 White Light Scanner from Steinbichler Optotechnik GmbH, Germany (Figure 3)**.

The white light scanner projected fringe (light) patterns on the bracket and the camera simultaneously captured the images, then advanced software algorithms triangulate and calculated the 3d-coordinates of numerous points spaced all over the surface of the bracket. The part of the bracket that was within the frame (illuminated zone with fringes) got scanned during a single measurement. For scanning the complete bracket several such measurements were carried out and merged together.

In our study we also had to go through a hybrid modelling procedure where parametric CAD Modelling and Rapid Surfacing had to be employed. This is due to the fact that the brackets were too small by engineering standards and it had a combination of freeform geometry as well as geometric shapes.

The following steps were followed for hybrid modelling:

i. Phase 1- The scan data was imported in software progran “Geomagic Studio”. Saving the bracket base all other regions were removed. This was exported as a **Standard for the Exchange of Product** (STEP) model data model to be used later.

ii. Phase 2 - The scan data was imported into Pro/Engineer Wildfire5 and the same was aligned using symmetry planes. Similarly, cuts and slots were added and gradually the entire bracket was modelled.

iii. Phase 3- The STEP model from Phase 1 was imported in the same file and this by default would form the base for the bracket, upon the upper half which was modelled in Phase 2. Corner radius was then added and the file was exported as IGS, STEP, or STL file, and this was used for method FEA.

Once the scanning was completed, the data was post-processed (includes steps like alignment, matching, decimation, and smoothing) and exported into a 3D point cloud/ triangle mesh which is either a Binary or ASCII (.**STL file**). The data is of high accuracy, and the repeatability of the White Light Scanners is between 1-5 microns. **(Figure 4)**

2) **PROCESSING STAGE:-** during this phase the software program **Ansys Workbench Version 11** was used, which imported models with 100% data transfer or with 0% data loss. Once the data of the brackets, tooth and its supporting structures was imported, the software performed automatic meshing with defined material properties. The software established contacts automatically and defined them as a bonded contact.**(Figure 5)**

This means that the wire was not deformed until it came in contact with the slot walls. Thus the wire mobility was restricted by the slot walls and the ligature, respectively. A frictional coefficient "μ" between the bracket and the wire of 0.2 was used. The bracket of the maxillary central right incisor was torqued from its neutral position by a total of 20 degrees[47] and the resultant forces were evaluated at 0mm, 4mm, 8mm and 12mm from the apex till the cervical region. The engagement angle that is is the amount of axial rotation that the wire is permitted to undergo before it contacts with the slot walls for the selected brackets was also evaluated.

3) **POST PROCESSING STAGE:-** the torque angle/torque moment values in the simulated movements were recorded by the FE software package, Ansys Workbench Version 11 and evaluation of the results was performed from the graphical, numerical and animated format **(Figure 6).**

Material	**Youngs Modulus**	**Poisons Ratio**
Tooth	2.00E+04	0.3
PDL	6.80E-01	0.49
Cortical Bone	1.40E+04	0.31
Cancellous Bone	2.50E+02	0.3
Bracket	2.14E+05	0.3
Wire	2.14E+05	0.3

Table 1: Material properties of Tooth, PDL, Bone, Bracket and Wire

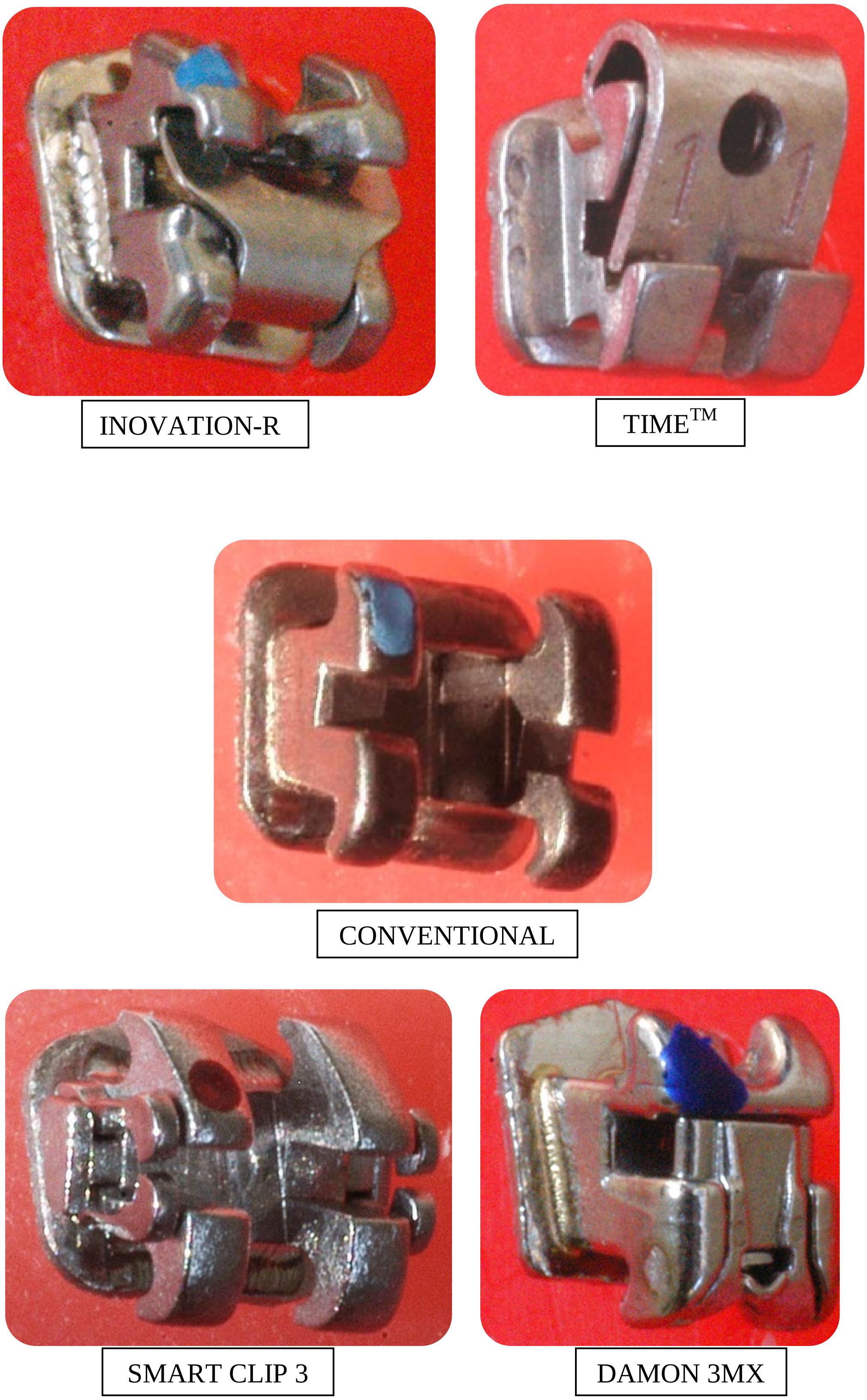
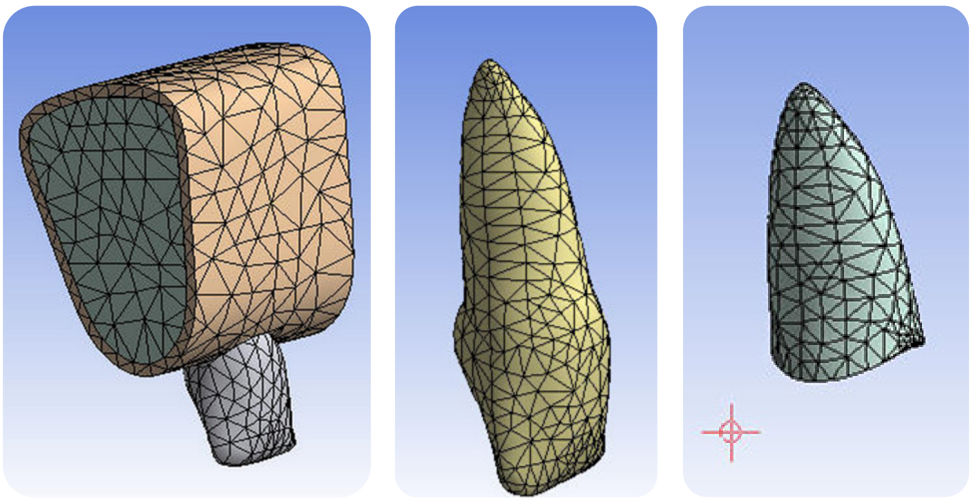


Figure 1: Brackets used in the study

(a) Computed Tomography Scanner (GE Healthcare Technologies - Lightspeed VCT)

(b) Computed Tomography scan of the maxilla



(c) 3 dimensional modelling of the tooth and its supporting structure

Figure 2: Pre-Processing Stage for the Tooth and its supporting structure

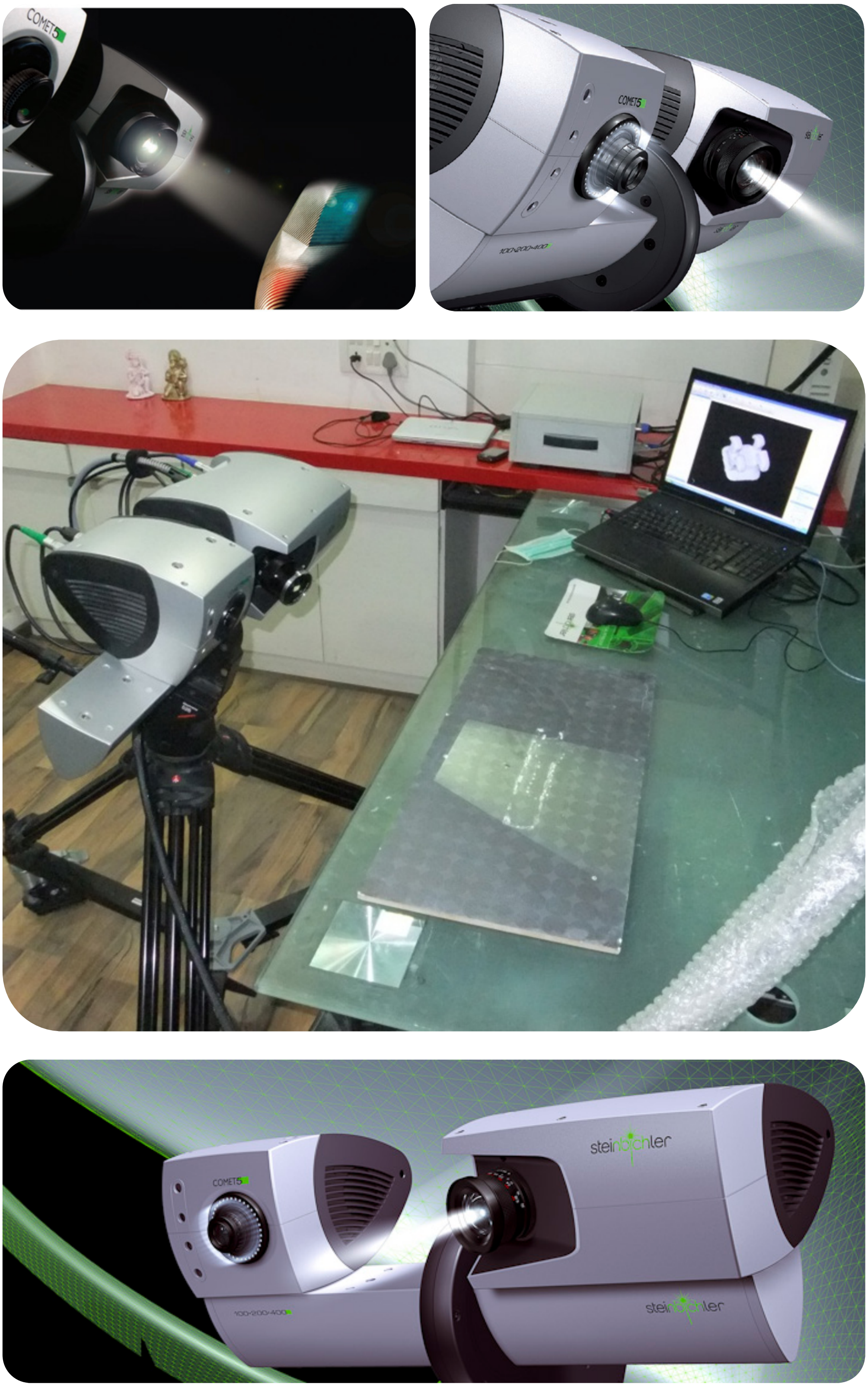


Figure 3: Comet 5 White Light Scanner (Steinbichler Optotechnik GmbH, Germany) used for scanning the brackets.

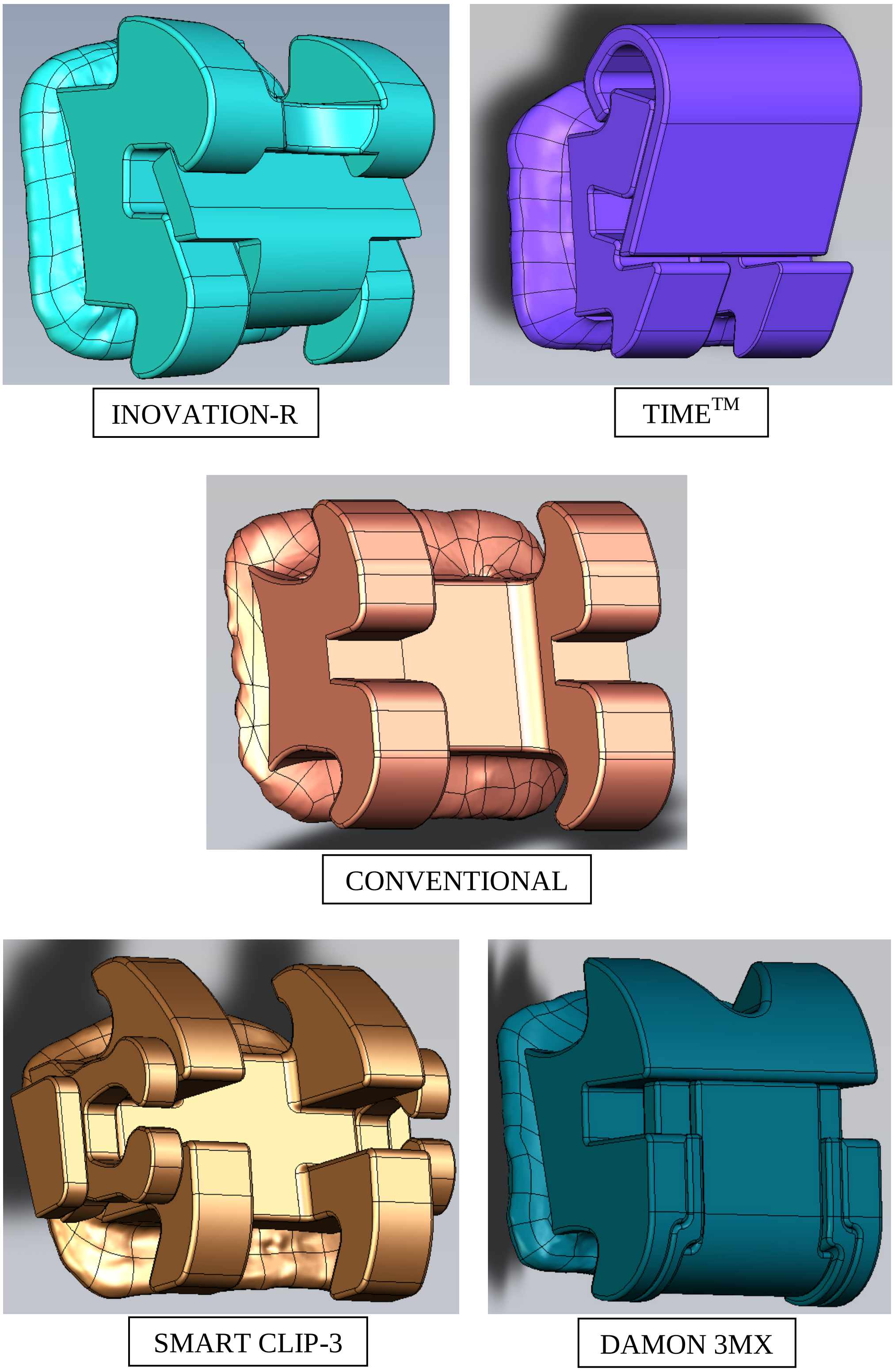


Figure 4: Pre-Processing Stage for the Brackets - 3 Dimensional models acquired from white light scanning.

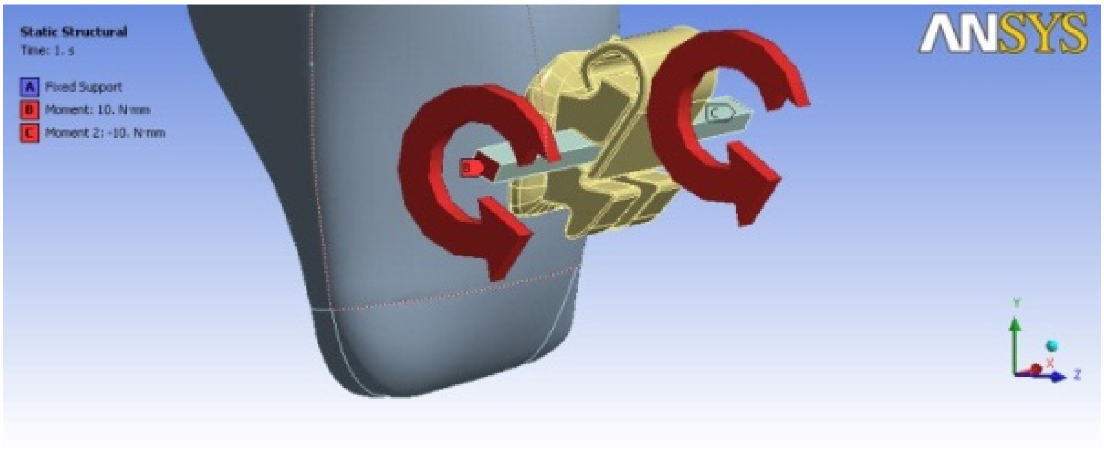


Figure 5: Processing Stage: 20 degrees of Torque was applied to the maxillary right central incisor

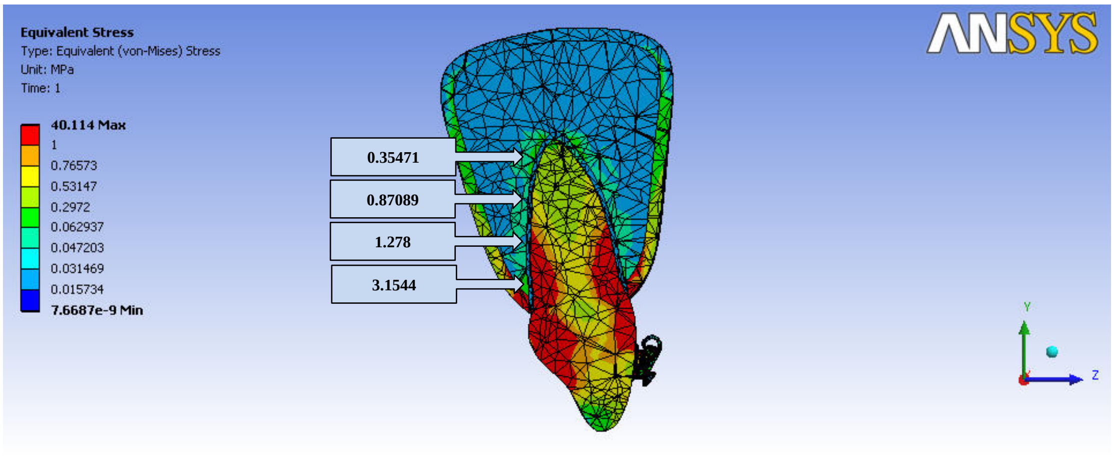


Figure 6: Post Processing Stage: Representation of the results in a colour coded manner

RESULTS

The values of the torque-moment delivered by various archwire dimensions onto the brackets and their resultant stress pattern on the root structure of the tooth and supporting periodontium were evaluated using the finite element analysis. They are shown in the spectrum of colours ranging from red (very high) to blue (lowest).

Resultant Force with 0.017 x 0.025-in S.S.Archwire: The maximum torquing moments were generated by the Conventional Bracket (41Nmm) followed by InOvation-R (34.6Nmm), Damon 3MX (34.3Nmm), Smart Clip-3 (32.5Nmm) and Time™ (32Nmm) bracket. **(Table 2, Graph 1)**

Resultant Force with 0.019 x 0.025-in S.S.Archwire: The maximum torquing moments were generated by the Conventional bracket (47.4Nmm) followed by InOvation-R (40Nmm), Damon 3MX (39.3Nmm), Smart Clip-3 (38.1Nmm) and Time™ (36.8Nmm) bracket. **(Table 2, Graph 2)**

Resultant Force with 0.021 x 0.025-in S.S.Archwire: The maximum torquing moments were generated by the Conventional bracket (79Nmm) followed by InOvation-R (64Nmm), Damon 3MX (63.8Nmm), Smart Clip3 (63.5Nmm) and Time™ (61.2Nmm) bracket. **(Table 2, Graph 3)**

Stress Pattern by Various Bracket-Archwire combinations on the Tooth and Periodontium at different levels: Using the values obtained from the finite element analysis, the forces for the simulated torque moments were plotted for each of the bracket-archwire combination used in the study **(Table 4)**. The results were consistent in all the groups tested.

The stress pattern values showed an increasing gradient from the apical third to the cervical region of the root surface when tested sequentially with S.S. wires, starting with 0.017 x 0.025-in, followed by 0.019 x 0.025-in, and finally 0.021 x 0.025-in (**Table 4)**.

Maximum Torquing Moments of brackets with variations in Archwire dimension: The graph summarizes the maximum torquing moments expressed by Conventional, InOvation-R, Time^{TM}, Smart Clip-3 and Damon3MX brackets with variations in archwire dimension.

The conventional bracket system consistently exhibited the maximum torquing moments with the three archwires tested (0.017x0.025, 0.019x0.025 and 0.021x0.025 inch wire). Within the self ligating group, the maximum torquing moments were expressed by InOvation-R followed by Damon3MX, Smart Clip-3 and Time^{TM} bracket **(Graph 4)**

Angle of engagement for different brackets: There was considerable variation in the engagement angle among the bracket systems assessed. The Conventional bracket showed the least engagement angle of 3º followed by InOvation-R (4.3º), Damon3MX (6º), Time^{TM} (7.1º) and the highest by Smart Clip-3 (7.9º). (**Table 3, Graph 5)**

TABLES

	Bracket	0.017 x 0.025	0.019 x 0.025	0.021 x 0.025
1	**InOvation-R**	34.6 Nmm	40 Nmm	64 Nmm
2	**Time^{TM}**	32 Nmm	36.8 Nmm	61.2 Nmm
3	**Smart Clip-3**	32.5 Nmm	38.1 Nmm	63.5 Nmm
4	**Damon 3MX**	34.3 Nmm	39.3 Nmm	63.8 Nmm
5	**Conventional**	41 Nmm	47.4 Nmm	79 Nmm

Table 2: Torque values of different bracket-archwire combinations

	Bracket	Wire	Engagement Angle (degrees)
1	**InOvation-R**	0.021x0.025	4.3
2	**Time^{TM}**	0.021x0.025	7.1
3	**SmartClip3**	0.021x0.025	7.9
4	**Damon3MX**	0.021x0.025	6
5	**Conventional**	0.021x0.025	3

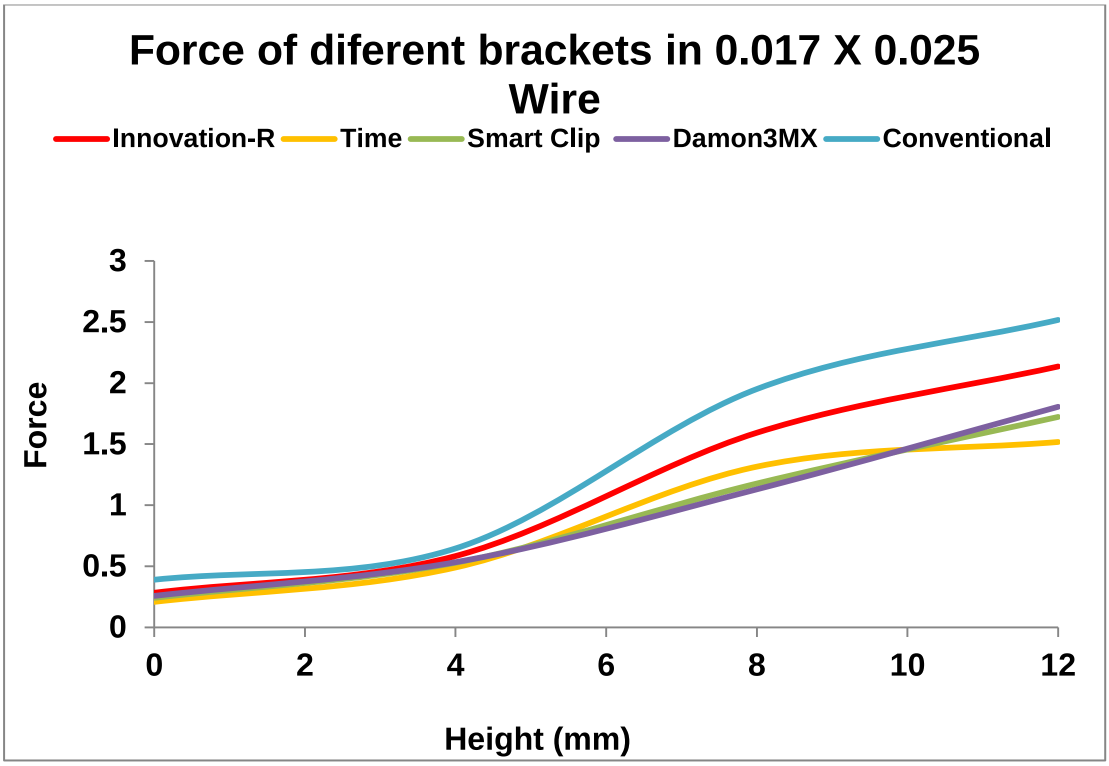
Table 3: Angle of Engagement for different bracket - archwire combinations

Bracket	Archwire	Height	Force
1)InOvation-R	0.017 x 0.025	0mm	0.34572
		4mm	0.5239
		8mm	0.95918
		12mm	1.8218
	0.019 x 0.025	0mm	0.28313
		4mm	0.58375
		8mm	1.5934
		12mm	2.1359
	0.021 x 0.025	0mm	0.4607
		4mm	0.92308
		8mm	1.5966
		12mm	3.3512
2)Time™	0.017 x 0.025	0mm	0.29834
		4mm	0.48038
		8mm	0.917467
		12mm	1.4707
	0.019 x 0.025	0mm	0.20859
		4mm	0.49008
		8mm	1.31567
		12mm	1.5178
	0.021 x 0.025	0mm	0.35471
		4mm	0.87089
		8mm	1.278
		12mm	3.1544
3)Smart Clip-3	0.017 x 0.025	0mm	0.25926
		4mm	0.43482
		8mm	0.92753
		12mm	1.6246
	0.019 x 0.025	0mm	0.24122
		4mm	0.52882
		8mm	1.1774
		12mm	1.72286
	0.021 x 0.025	0mm	0.46616
		4mm	0.90113
		8mm	1.3939
		12mm	3.4628

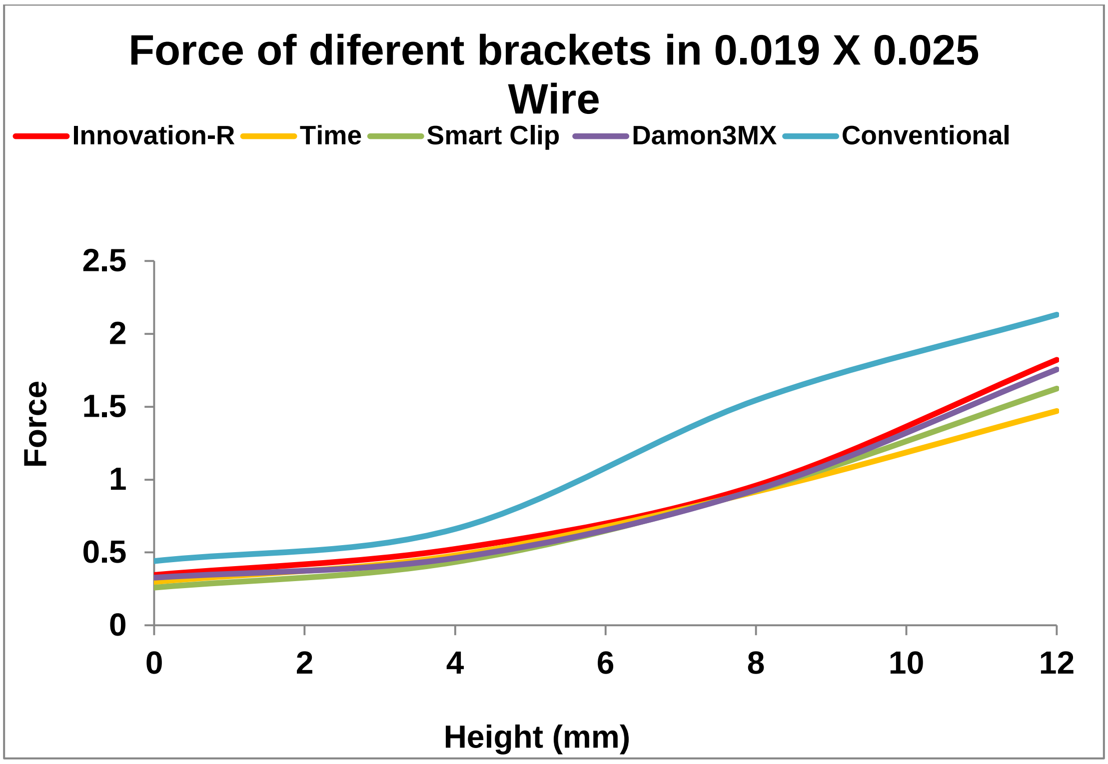
Table 4: Stress Values at 0, 4, 8 and 12mm induced by various bracket-archwire combinations on the root surface of the tooth and its supporting periodontium.

4) Damon 3MX	0.017 x 0.025	0mm	0.32737
		4mm	0.46204
		8mm	0.930193
		12mm	1.75588
	0.019 x 0.025	0mm	0.25748
		4mm	0.53329
		8mm	1.13
		12mm	1.8058
	0.021 x 0.025	0mm	0.46617
		4mm	0.91332
		8mm	1.5963
		12mm	3.4998
5)Conventional	0.017 x 0.025	0mm	0.44112
		4mm	0.66171
		8mm	1.54483
		12mm	2.1316
	0.019 x 0.025	0mm	0.38994
		4mm	0.64615
		8mm	1.95121
		12mm	2.5171
	0.021 x 0.025	0mm	0.47553
		4mm	1.19607
		8mm	1.8501
		12mm	4.3597

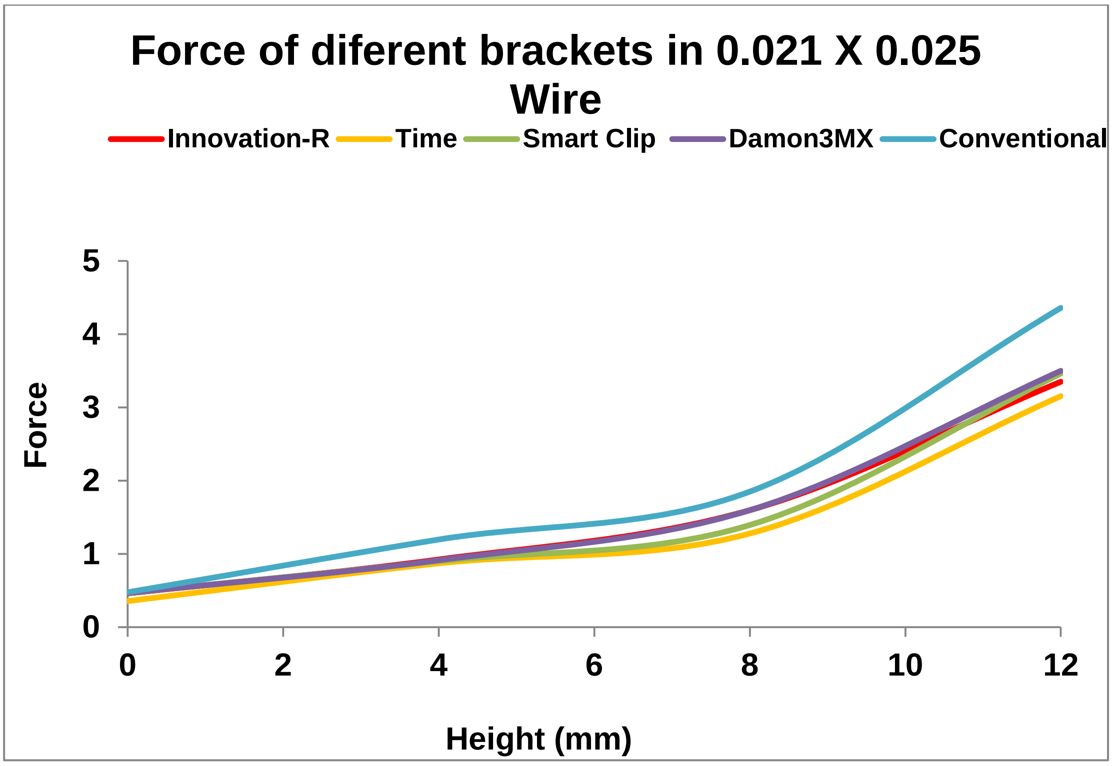
Table 4: Stress Values at 0, 4, 8 and 12mm induced by various bracket-archwire combinations on the root surface of the tooth and its supporting periodontium.



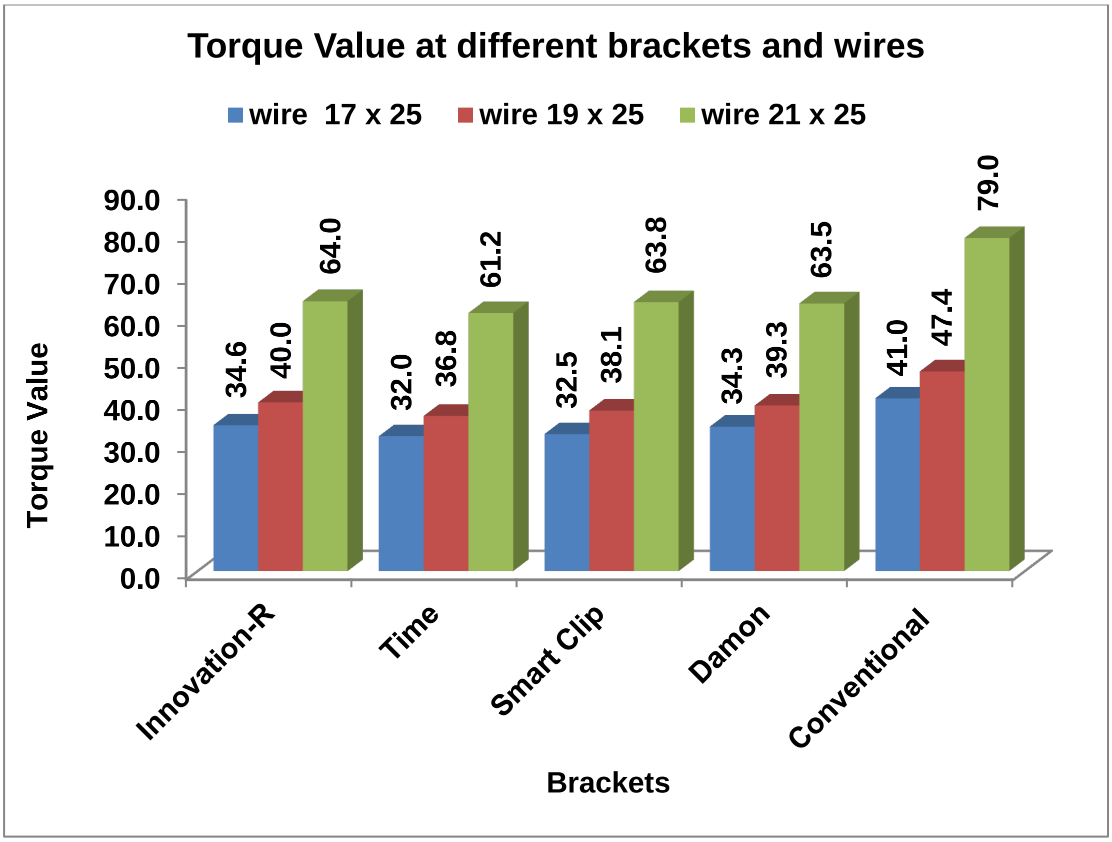
Graph 1: Stress Pattern curves of the simulated torque moments with "0.017x0.025 inch" S.S.Archwire for various brackets.



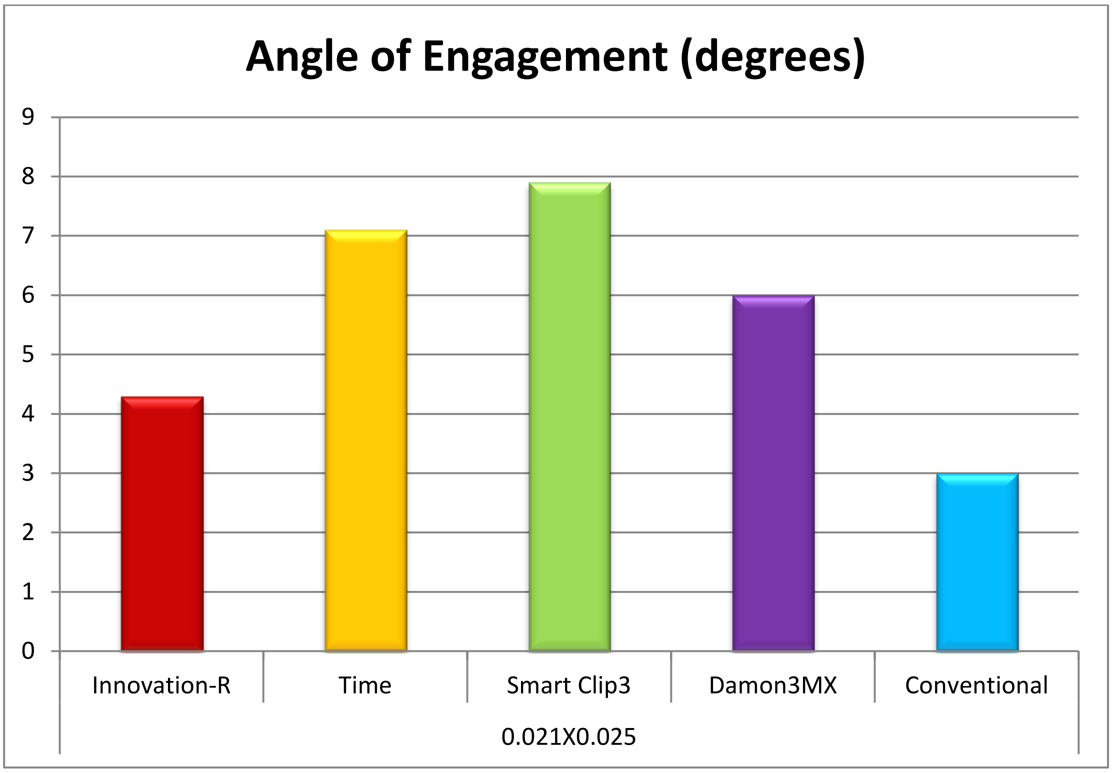
Graph 2: Moment-torque activation curves of the simulated torque moments with "0.019x0.025 inch" S.S.Archwire for various brackets.



Graph 3: Moment-torque activation curves of the simulated torque moments with "0.021x0.025 inch" S.S.Archwire for various brackets.

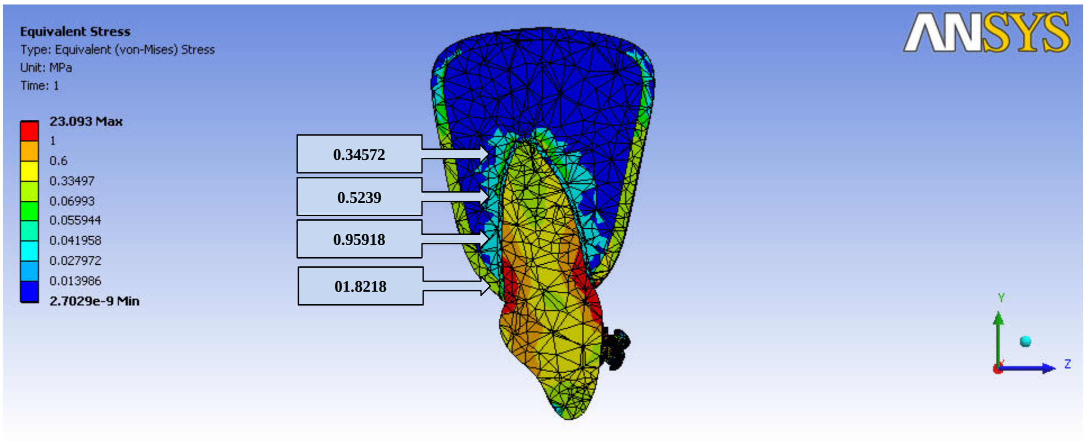


Graph 4: Maximum Torquing Moments of brackets with variations in Archwire dimension

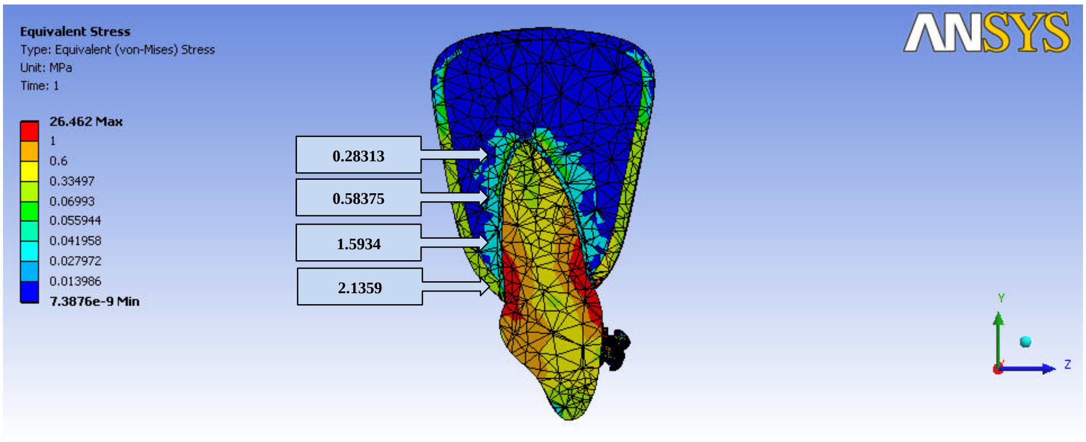


Graph 5: Engagement angle of different brackets.

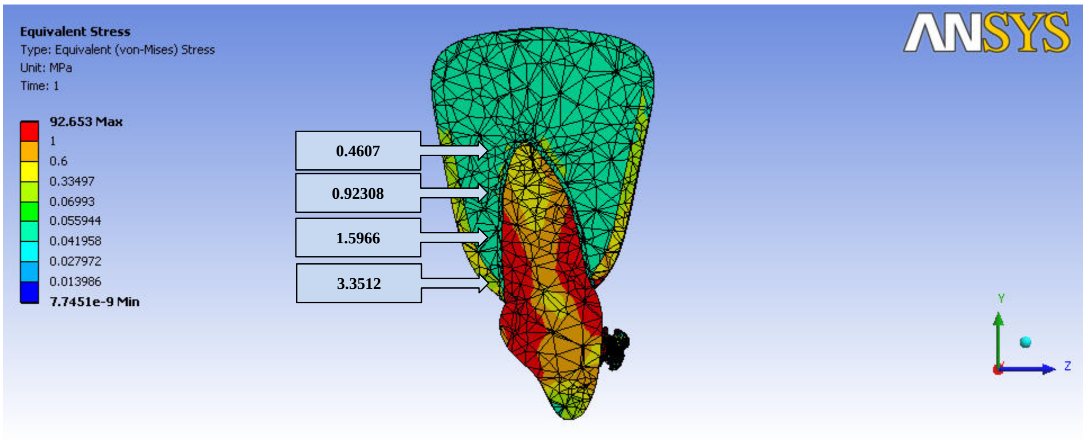
Figure 7: STRESS PATTERN ON THE TOOTH AND PERIODONTIUM BY "INOVATION-R" BRACKET



(a) 0.017 x 0.025" Stainless Steel Archwire

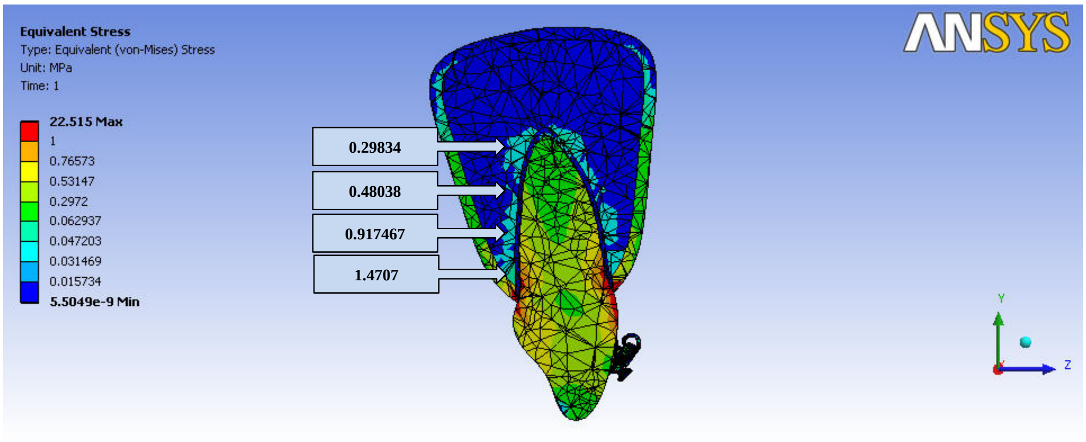


(b) 0.019 x 0.025" Stainless Steel Archwire

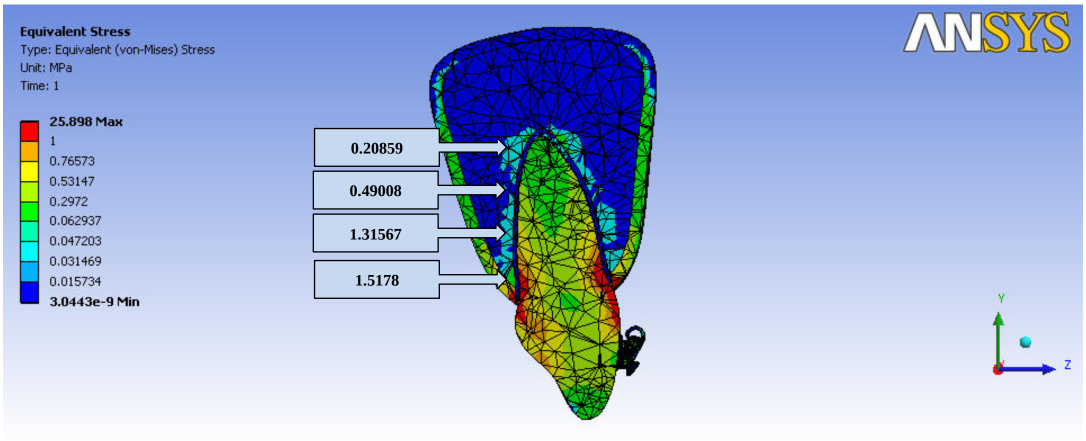


(c) 0.021 x 0.025" Stainless Steel Archwire

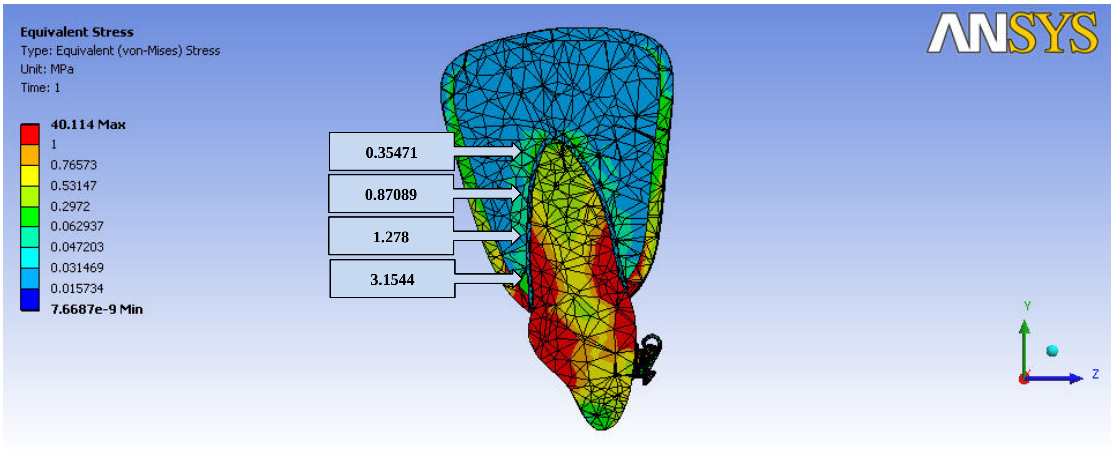
Figure 8: STRESS PATTERN ON THE TOOTH AND PERIODONTIUM BY "TIME™" BRACKET



(a) 0.017 x 0.025" Stainless Steel Archwire

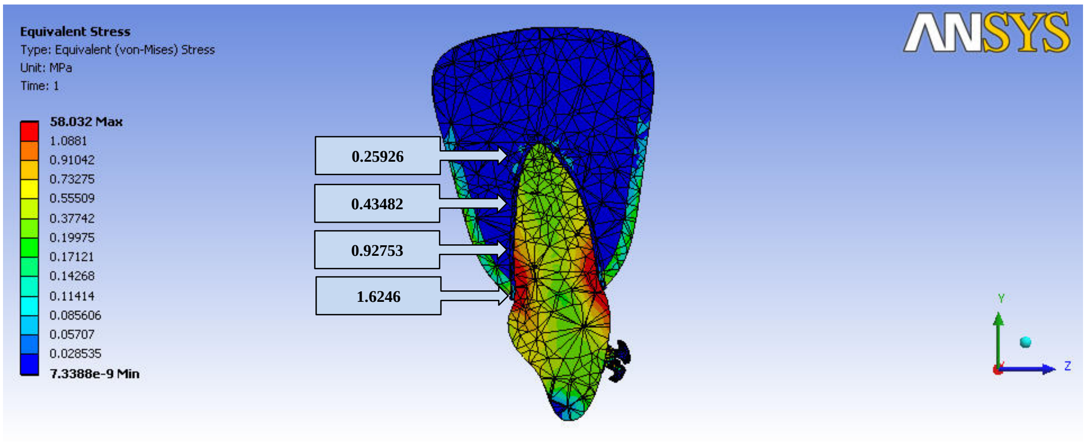


(b) 0.019 x 0.025" Stainless Steel Archwire

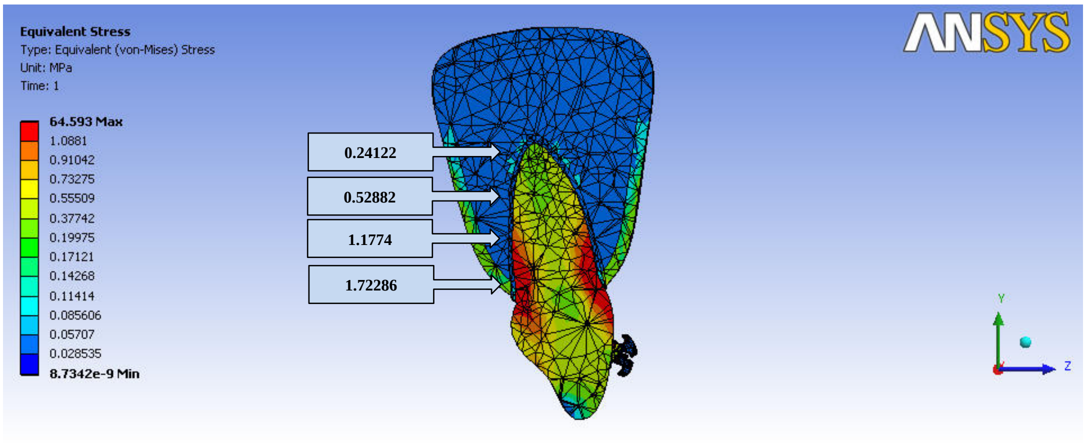


(c) 0.021 x 0.025" Stainless Steel Archwire

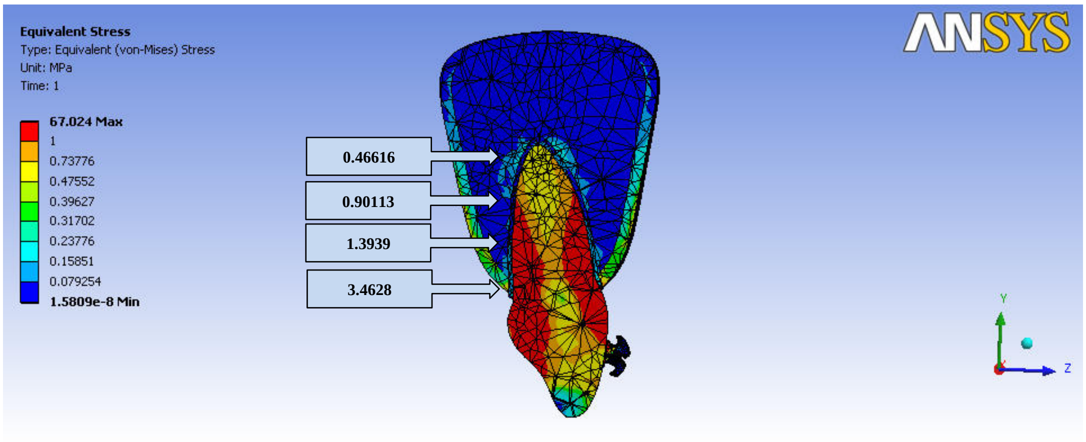
Figure 9: STRESS PATTERN ON THE TOOTH AND PERIODONTIUM BY "SMART CLIP-3" BRACKET



(a) 0.017 x 0.025" Stainless Steel Archwire

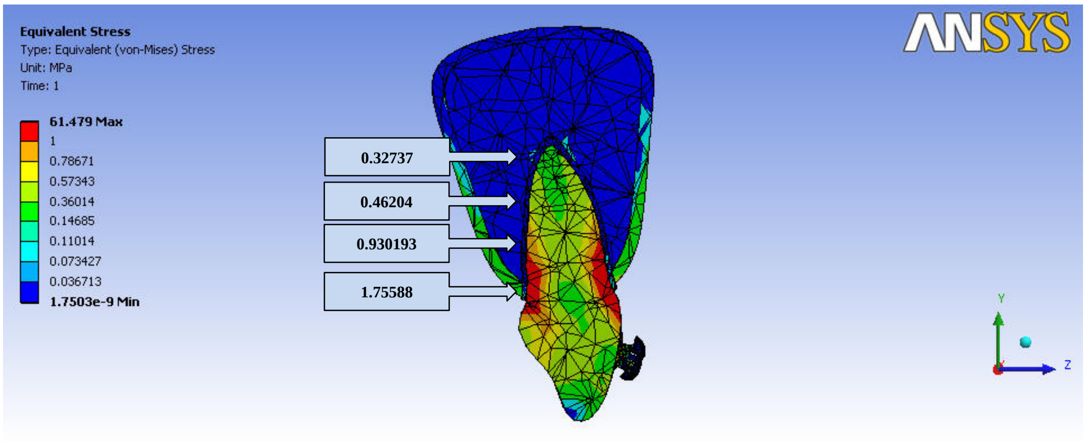


(b) 0.019 x 0.025" Stainless Steel Archwire

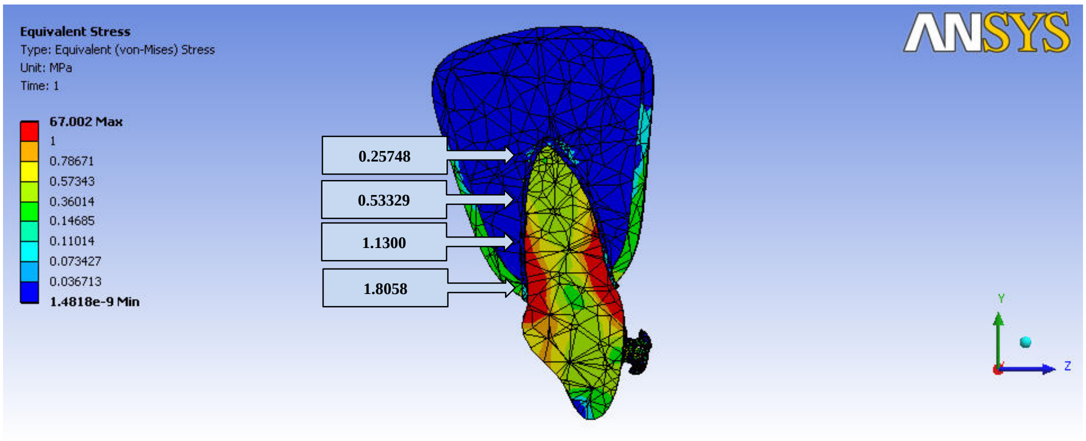


(c) 0.021 x 0.025" Stainless Steel Archwire

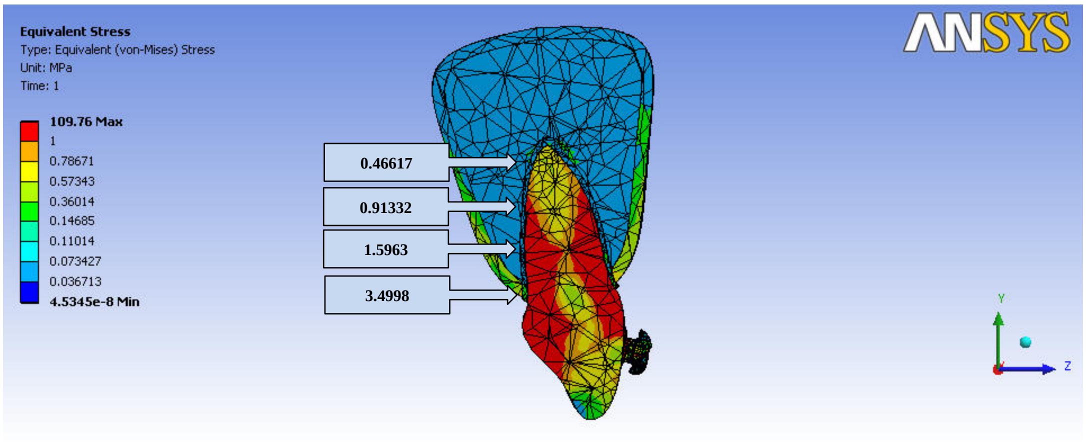
Figure 10: STRESS PATTERN ON THE TOOTH AND PERIODONTIUM BY "DAMON 3MX" BRACKET



(a) 0.017 x 0.025" Stainless Steel Archwire

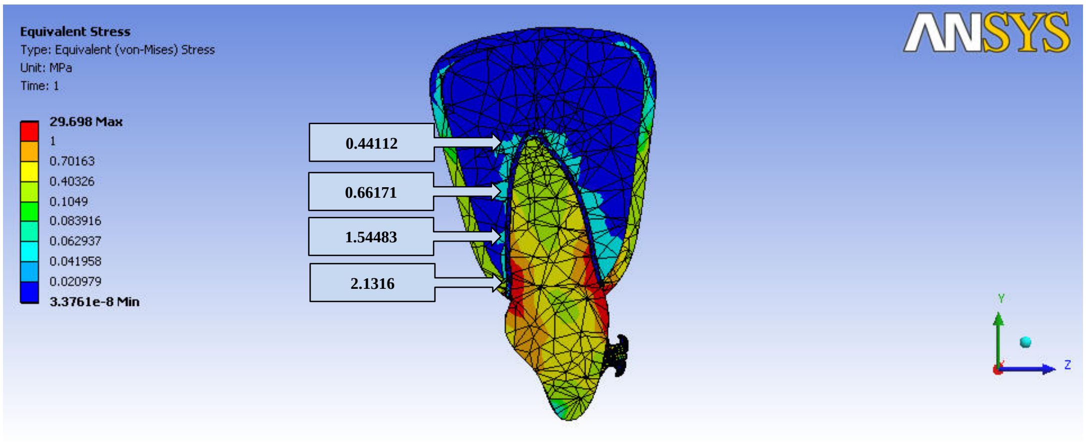


(b) 0.019 x 0.025" Stainless Steel Archwire

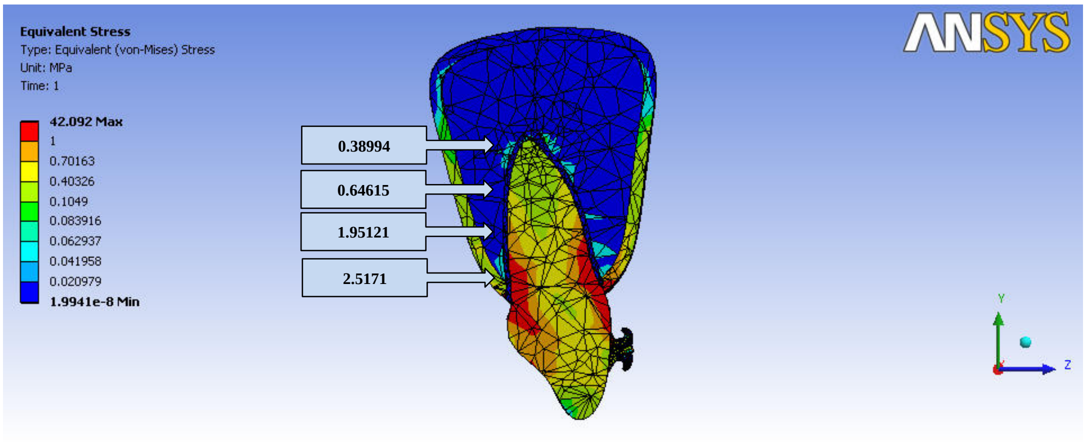


(c) 0.021 x 0.025" Stainless Steel Archwire

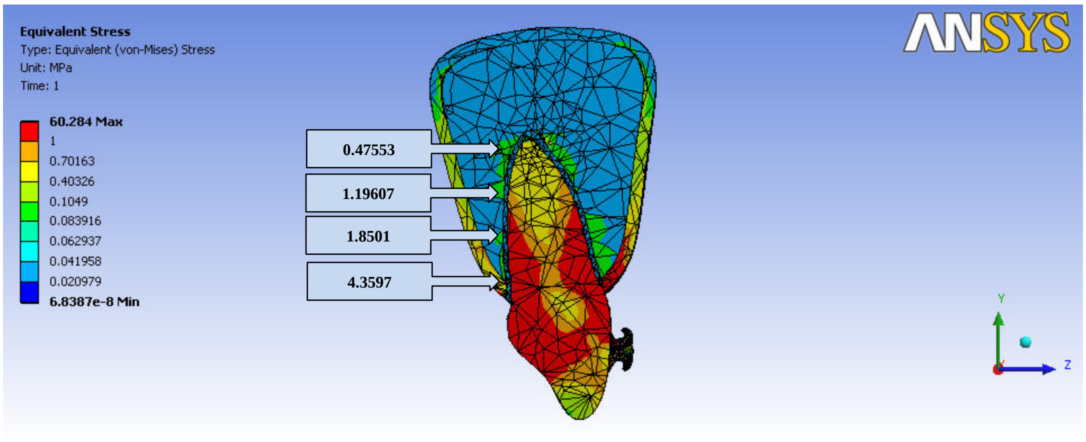
Figure 11: STRESS PATTERN ON THE TOOTH AND PERIODONTIUM BY "CONVENTIONAL" BRACKET



(a) 0.017 x 0.025" Stainless Steel Archwire



(b) 0.019 x 0.025" Stainless Steel Archwire



0.021 x 0.025" Stainless Steel Archwire

DISCUSSION

The specialty of orthodontics has continued to evolve since its advent in the early 20th century. Changes in treatment philosophy, mechanics, and appliances have helped shape our understanding of orthodontic tooth movement.

A major shift in orthodontics occurred when **Andrews**[3] introduced the “straight wire appliance”. Instead of bending wires to place teeth in the proper orientation with an edgewise bracket, the Andrews appliance had the tip and torque values built into the brackets commonly known as the “appliance prescription”. In theory, these pre-adjusted brackets eliminated the need to repeatedly bend first, second, and third order bends each time the patient progressed to the next wire. The straight wire appliance revolutionized orthodontics by making the bracket much more efficient. Since then, many orthodontic companies have developed their own bracket systems with specific prescriptions, treatment philosophies, and mechanics.

In the recent years, there has been a boost in the manufacturing and release of self-ligating appliances with active or passive ligation modes, leading to entice more clinicians due to their proclaimed time-saving ligation mode and the potential alterations in the load and moment expression during mechanotherapy. Some of these systems seem to present reduced friction, however their torquing characteristics remain uncertain.

Although there is a conundrum concerning which self ligating bracket to set into practice, the issue of active clip or passive slide/clip is a major focus of the controversy.

Self-ligating brackets are broadly classified into Active and Passive self-ligating brackets;

1) **Active self-ligating brackets:** Active brackets consist of a spring clip which comes in contact with the arch wire when engaged. Automatic seating of either a round or a rectangular archwire at the base of the slot is responsible for the light, continuous force[31]. The spring clip stores energy to press against the archwire for greater torque control.[6] In the active self-ligating system, more friction is produced as a result of the clip pressing against the archwire.[62]
2) **Passive self-ligating brackets:** In passive self-ligating bracket the slot is transformed into a tube by means of a labial "fourth wall" that does not contact the archwire.[21] The full expression of bracket properties is achieved only when higher dimensional wires are used and the torque control is efficiently achieved only by using larger rectangular archwires.[67,84]

Torque as described by Rauch[88], is a moment generated by the torsion of a rectangular wire in the bracket slot. Torque can also be defined from a mechanical and a clinical point of view. Mechanically, it refers to the twisting of a structure about its longitudinal axis, resulting in an angle of twist. Torque is a shear-based moment that causes rotation. Clinically, in orthodontics, it represents the buccopalatal crown/root inclination of a tooth, and it is an orthodontic adaptation used to describe rotation around the x-axis. When applied in an orthodontic archwire/bracket interaction, it describes the activation generated by twisting an archwire in a bracket slot[114].

Correct buccolingual inclination of anterior teeth is considered essential for providing good occlusal relationships in orthodontic treatment. Inclination of the maxillary anterior teeth is particularly critical in establishing an esthetic smile line, proper anterior guidance, and a Class I canine and molar relationship. Undertorqued maxillary anterior teeth affect the arch length and the space requirements. It has been shown that for every 5° of anterior inclination, about 1 mm of arch length is generated[78]. Undertorqued posterior teeth have a constricting effect on the maxillary arch, since they do not allow appropriate cusp to fossa relationships between the maxillary and mandibular teeth[113].

Tip, in–out, and rotation control have become highly uniform in all current and popular appliance prescriptions. Torque, on the other hand, is available in a variety of ranges. This would occur due to several factors: mechanical side-effects, morphological differences in the buccal faces of teeth, changes in the position of the brackets, different methods of bracket manufacturing and orthodontic wires, the play between the wire and the bracket slot, variations in the bracket designs, properties of the materials constituting the brackets and wires and differences between the value of the torque informed by the manufacturer and the real value of the torque of the brackets[96].

Numerous studies have measured the torque characteristics of the bracket systems by a number of testing apparatus.

Gmyrek[39], **Harzer**[43], and **Morina**[76] used the Orthodontic Measurement and Simulation System (**OMSS**[10]) to measure the maximum torquing moment of self ligating bracket systems. The major components of the OMSS are the two force – moment sensors capable of measuring forces and moments simultaneously in all three planes of space.

Badawi et al[6] developed a novel apparatus with a digital inclinometer to evaluate the torque expression of self ligating brackets. Torque was evaluated as the wire was twisted, all the other forces and moments were set to zero by device alignment. Vertical and horizontal alignment was maintained between the wire and the bracket during this process.

However, these methods are quite cumbersome and depend on extensive instrumentation and further they fail to graphically display the changes for the clinician to appreciate.

Finite Element Analysis (FEA) is a powerful computer-simulation tool for solving stress-strain problems in the mechanics of solids and structures in engineering. The study of orthodontic biomechanics requires the understanding of the stress and strain induced by orthodontic forces. Finite element analyses (FEA) offer a means of determining stresses in tooth, ligament, and bone structures for a broad range of orthodontic loading conditions.

Thus, Finite Element Method is considered a superior method for determining stress distribution patterns and resultant force on structures of complex designs and known material properties.

To the best of our knowledge, finite element analysis to investigate the torque expression of self ligating on the tooth and periodontium is very scant.

Accordingly, this study was designed to investigate the torque expression of different self-ligating brackets and arch wire combinations on the tooth and its supporting structures implicating finite element method.

In the present study self-ligating brackets were divided into two groups- **Active clip type** [InOvation-R (GAC Intl, NY) & Time (American Orthodontics, USA)] and **Passive clip type** [Damon3MX (Ormco Orthodontics, California) & SmartClip3 (3M Unitek, United States)] whereas preadjusted twin bracket [Ovation (GAC Dentsply,USA)] served as control. These brackets were tested for their torque proficiency offered to stainless steel archwires. Three types of Stainless steel wires with varying dimensions were used, 0.017x0.025-in, 0.019x 0.025-in & 0.021 X 0.025-in wires.

The upper central incisor was preferred for this study because the torque control of upper central incisor is considered of paramount importance in clinical situation.

A 3 dimensional model of the right maxillary central incisor and its supporting structures was generated from a Computed Tomography **(CT)** scan by a **Computer Aided Designing/Computer Aided Engineering** (CAD/CAE) program. The creation of a periodontal ligament with a thickness of 0.25 mm was performed due to the impossibility to define this structure from CT images (pixel size = 0.273 mm).

The brackets used in the study were scanned and 3 dimensional models and designed with **Comet 5** k **White Light Scanner from Steinbichler Optotechnik GmbH, Germany**. The white light scanner projected fringe (light) patterns on the bracket and the camera simultaneously captured the images, then advanced software algorithms triangulated and calculated the 3d-coordinates of numerous points spaced all over the surface of the bracket. The part of the bracket that was within the frame (illuminated zone with fringes) got scanned during a single measurement. For scanning the complete bracket several such measurements were carried out and merged together.

The software program, **Ansys Workbench 11**, was used for the study which can import models with 100% data transfer or with 0% data loss. Once the data was imported the software performed an automatic meshing with defined material properties. The software established contacts automatically and defined them as bonded contact.

This means that the wire was not deformed until it came in contact with the slot walls. Thus the wire mobility was restricted by the slot walls and the ligature, respectively. A frictional coefficient "μ" between the bracket and the wire of 0.2 was used. The bracket of the maxillary central right incisor was rotated from the neutral position by a total of 20 degrees and the resultant forces were evaluated at 0mm, 4mm, 8mm and 12mm from the apex to cervical region of the root. The engagement angle for the selected brackets was also evaluated.

The results of the present study indicate the maximum torque values were consistently exhibited with the conventional ligation system when compared to the self ligating bracket systems. Amongst the self ligating brackets, InOvation-R exhibited the maximum torque values followed by Damon 3MX, Smart Clip-3, and the least torque values were exhibited by the Time^{TM} self ligating bracket system **(Table 2, Graph 4).** Similar findings were observed in the study conducted by **Morina et al**.

In the InOvation-R bracket, the slot has a short gingival horizontal wall of 0.0195-inches and a conventional occlusal horizontal wall of 0.0285 inches which lets the spring clip invade the slot depth (**Figure 1)**. It was this unique feature in the design of the bracket that helped in establishing more interaction between the archwire and the bracket, thereby improving the torque expression. This finding correlates with the earlier findings of **Major et al**[65]**.**

The Damon3MX self ligating brackets consist of a sturdy passive slide which upon closing forms the fourth wall of the bracket slot. The full expression of bracket properties is achieved only when higher dimensional wires are used and the torque control is efficiently achieved only by using larger rectangular archwires.[21]

The Smart Clip-3 self ligating brackets consist of nickel-titanium spring clips mesial and distal to the tie wings to capture the archwire inside the slot. This characteristic feature of the bracket that would have probably resulted in lesser amount of torque expression due to elastic deformation of the clip when interacting with larger archwire dimensions. This was in accordance with results published by **Badawi et al**[6]**.**

In the Time self ligating bracket system, the active clip establishes a contact with the rest stop of the bracket first unless a full size archwire is employed. Due to its rigid nature, the clip itself possesses no elastic energy. In this respect, once closed, the bracket behaves similarly to the passive self ligating bracket, assuming the archwire is sitting passively within the bracket slot. As a consequence of this clip design, the prescribed torque available was reduced**.** Similar results were revealed by **Budd et al**[12]**.**

The maximum torque values were consistently exhibited with the conventional ligation system. This is credited to the robust design of the bracket and better engagement of the archwire within the slot by the stainless steel ligature ties.

The torquing values displayed an escalation as the archwire dimensions were stepped up from 0.017 x 0.025 inches to 0.021 x 0.025 inches in all the bracket groups examined. This signifies the importance of archwire dimension in torque expression.

In the present study, the angle of engagement for all the bracket groups was also measured against the 0.021x0.025 inch archwire.

The angle of engagement is the amount of axial rotation that the wire is permitted to undergo before it contacts with the slot walls. This angle was selected for evaluating the torque-play between the different bracket systems. After engaging the archwire in the bracket slot, the degree of torque expressed depends on the surface area of the bracket slot contacted by the archwire. Therefore, degree of angle of engagement is inversely proportional to the torque expressed by the brackets.

In the present study, the conventional bracket exhibited the least angle of engagement when compared to the self ligating bracket systems. Amongst the self ligating brackets the least angle of engagement was exhibited by InOvation-R, followed by Damon 3MX, Time and highest angle of engagement by Smart Clip-3 self ligating system. These findings are concurring with those of **Fischer Brandies et al**[26].

The InOvation-R bracket presented the least angle of engagement (4.3º). This was probably due to the encroachment of the slot by the active spring clip which helped in establishing enhanced interaction between the archwire and the bracket slot and increasing the torque expression.

The Damon 3MX bracket presented an angle of engagement of 6º. This was the amount of play exhibited by the archwire within the bracket slot before it contacted the walls of the bracket. Thus, while using the Damon 3MX bracket it must be kept in mind that to achieve proper torque control larger archwire dimensions that fill the slot must be used. This was in accordance with the study conducted by **Huang et al[47].**

The Time™ bracket presented an engagement angle of 7.1º. The increase in the play of the wire is attributed to the design of the active spring clip which actually rests passively on the rest stop of the bracket instead of encroaching it.

The Smart Clip 3 bracket presented the highest angle of engagement of 7.9º. This could probably be due to the inability of the nickel-titanium clips on the mesial and distal tie wings of the bracket to engage the archwire at earlier stages of torsion.

The conventional bracket system exhibited the least angle of engagement of 3 º. This is the prime advantage of conventional bracket system compared with all the self ligating bracket systems evaluated. The stainless steel ligature tie around the bracket wings reduce the amount of torque play and thus enhance the torque expressed by the bracket. This was in accordance with the studies conducted by **Badawi et al[6]**, **Morina et al[65]**, and **Huang et al[47].**

Consequently, the present study demonstrated that the angle of engagement plays an important role in the torque expressed by the various brackets. This was established by the results which revealed higher torquing moments for the brackets which had lower engagement angles as observed in the Conventional, InOvation-R and Damon 3MX bracket system. However, in spite of this fact, the Time™ self ligating bracket which had lower engagement angle as compared to the Smart Clip-3 bracket failed to express higher torque values when compared to it. This may be owed to the incapability of the active clip in the Time™ bracket to maintain the interaction between the archwire and bracket slot at higher torquing forces.

However, the results of this study are not in agreement with a recent clinical investigation by **Pandis et al**[83], which examined maxillary central incisor inclination with conventional and Damon brackets, and reported that there was no significant difference between the torque of incisors between the two appliances. However, the mechanotherapy used in that investigation greatly influenced the torque expression of the appliances since the use of rectangular NiTi reverse curve of Spee archwires, which are torqued more than 20 degrees, may cancel out any appliance variability in expressing torque.

The findings of the present study lay credence to the importance of bracket slot design and full slot archwire engagement and that angle of engagement is a parameter of clinical importance because it allows us to select a proper archwire dimension to effectively express the desired torquing moment.

FEM may give results with a reasonable degree of accuracy, but this approach has certain limitations. The accuracy of the analysis is dependant on the modelling of structures as closely as possible to the actual. However, a certain amount of approximation manifested chiefly in terms of type and number of arrangement of elements is inevitable in complex designs. Apart from this, one must be aware of the assumption used in the formulation, material characterization, nature of boundary conditions and the representations of loads. FEA has also failed to incorporate the time-dependent changes exhibited by various materials and their effects on the biological tissues. All these factors affect the validity of the results.

Thus, clinical trials are necessary to evaluate the in-vivo effects of the torque expression by different bracket-archwire combinations.

Further studies evaluating torque expression of self ligation brackets need to be carried out in order to:

1. Calculate the ideal amount of force required to deliver adequate torque by various self ligating bracket systems.
2. Estimate torquing moments with variation in archwire alloy.
3. Evaluation of torque expression on all maxillary anterior teeth simultaneously.

SUMMARY AND CONCLUSION

This FEM study was carried out to investigate the torque expression of different self-ligating brackets and arch wire combinations on the tooth and its supporting structures.

Two Active (Innovation-R and Time) and Passive (Smart Clip 3 and Damon3MX) self-ligating bracket systems were selected and one conventional (Ovation) bracket system served as control. Upper Right Central Incisor Stainless Steel Roth Prescription bracket with slot dimension of 0.022 x 0.028 inches was used in all the groups. The brackets were tested against three S.S. archwire dimensions (0.017x0.025, 0.019x0.025 and 0.021x0.025 inches).

A 3-dimensional model of the right maxillary central incisor and its supporting structures was generated from a Computed Tomography scan of a dry human skull by a CAD/CAE program. The brackets were scanned and 3-dimensional models were designed with Comet 5 White Light Scanner.

The close geometric diagram for the bracket, tooth and its supporting structures was prepared using Ansys Workbench Version 11. The bracket of the maxillary central right incisor was rotated from the neutral position by a total of 20 degrees.

The angle of engagement and the resultant forces (stress concentration) were evaluated at 0mm, 4mm, 8mm and 12mm from the apex for different archwire-bracket combinations were recorded using the same software.

Based on the finite element analysis, the following conclusions were drawn:-

(a) The maximum torque values were found with the conventional ligation system. These values exhibited an escalation as the archwire dimensions were stepped up from 0.017 x 0.025 inches to 0.021 x 0.025 inches. Among the self ligating bracket systems tested, Innovation –R provided superior torque values followed by Damon3MX, Smart Clip3 and least values were observed for the Time self ligating bracket system. Therefore, the torque-moment behavior is determined by archwire dimension and the design of the bracket.

(b) The least angle of engagement was observed for the Conventional bracket system followed by InOvation-R, Damon, Time and Smart Clip. Overall, this study indicates that the engagement angle is clinically significant, and is affected by archwire dimension, as well as by bracket slot dimension.

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