University of Alberta

Torque Expression and Bracket Deformation of the Orthos and Orthos Ti Orthodontic Bracket

by

Dr. Ryan A. Lacoursière

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Examining Committee

Dr. Paul W. Major, Dentistry

Dr. David Nobes, Mechanical Engineering

Dr. Giseon Heo, Dentistry

Dr. Jason Carey, Mechanical Engineering

Dedication

To my phenomenal wife who has provided me with such loving support and care throughout the orthodontic program. With her and my two wonderful children by my side, conquering the numerous challenges and frustrations that arose during this arduous process became a reality. Thank you so very much and you are forever my love.

To my parents whose determination and hard work ethic obviously imprinted upon me and provided me with the desire to always give 110% as they have throughout their lives.

Abstract

Deformation of the orthodontic bracket upon wire engagement may result in torque dissipation within the bracket rather than transmission to the tooth and its supporting structures. The purpose of this study was: 1) to quantify the amount of torque expression and 2) to quantify any deformation as a result of such increasing torque expression.

Digital image correlation is an accurate means of analyzing an orthodontic bracket's structural response to an applied moment of force created through archwire rotation in vitro.

A sample of 30 Orthos and 30 Orthos Ti brackets were tested with custom software using a novel application of digital image correlation.

OrthoTi brackets produce significantly greater amounts of torque from 48° to 24° returning to a neutral position. Titanium brackets exhibit greater variance in torque expression but less variance in deformation. Orthos stainless steel brackets exhibit significantly greater amounts of bracket slot deformation in comparison to the OrthosTi.

Acknowledgements

I thank the good the Lord for providing me with the opportunity to further my education and my personal development. In addition I wish to thank my beautiful family that I have been blessed with. Without them at my side throughout this roller coaster referred to as graduate studies, I surely could have become easily derailed. Furthermore, I thank my classmates, whom have become some of my closest and dearest friends. I dare say that you were the greatest reward I received from this program.

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My understanding regarding the use of imaging to evaluate and measure the effects of a force upon a material object, such as the orthodontic bracket is due to the efforts of Dr. Nobes and Dr. Carey. I have thoroughly enjoyed working with you in this project.

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List of Abbreviations

rtn - return to neutral

ss - stainless steel

SD – standard deviation

mm – millimeter

Ti – titanium

Nmm - Newton millimeters

OMSS – orthodontic measurement and simulation system

CCD – charge doupled device

NiTi – nickel titanium

β-Ti – beta titanium

MIM – metal injection moulding

VH = Vicker's Hardness

MBT – McLaughlin, Bennett, Trevisi

CMOS – complementary metal oxide semiconductor

bpp – bits per pixel

AISI – American Iron and Steel Institute

ANOVA – analysis of variance

ANCOVA – analysis of covariance

Chapter 1: INTRODUCTION and LITERATURE REVIEW

1.1 INTRODUCTION

Although being one of the most crucial elements of the edgewise appliance, torque is still one of the least understood fundamentals in orthodontic mechanics. With the predominant use of straightwire mechanics the importance of understanding torque and its expression is imperative to clinical efficiency and case proficiency. One can define torque in two manners as is relates to the orthodontist. The first describing it as the buccolingual root inclination of the tooth; the other being the force moment generated by torsion of a rectangular archwire in the rectangular bracket slot.[14] If the bracket either deforms in an elastic (temporary) or plastic (permanent) manner upon wire engagement, the magnitude of torque expression will be less than anticipated.

Torque expression is dependent upon slot and wire dimensions, wire properties, bracket properties, interbracket distance, and the degrees of wire rotation (twist) relative to the bracket slot.[2,6,7,11,24,25,26,27,28,29,30,31]
The vertical positioning of the bracket on the labial surface of the tooth [2,4,13] and the morphology of the dentition as it relates to the angle of the long axis between the root and crown the tooth[12] will influence the relative rotation of the wire relative to the bracket slot,
The optimal amount of force moment required to rotate teeth in a buccal/lingual manner has yet to be proven but a range of 5 to 20 Nmm has been published as being clinically effective.[6]

Few studies at present have critically analyzed deformation of the bracket structure in response to a torquing moment generated from the twisting of the wire within the bracket slot. Those that have focused primarily upon esthetic brackets of either polycarbonate or ceramic composition, rather than metal brackets and/or have been confined to analysis of the bracket structure regarding plastic change only.[16,21] The first objective of the present study was to construct a device that would measure torque expression concomitantly with the ability to record through digital imagery, change in bracket slot dimensions. The second study objective was to measure torque expression and bracket deformation (change in slot width dimension) for a stainless steel and titanium bracket of similar design.

1.2 Problem Statement

Although the profession of orthodontics has been inundated with advances in technology regarding diagnostic equipment that has undergone significant testing and validation; it appears as though much is taken for granted regarding the current and new materials used, especially in the area of brackets and wires. Few studies are published that test the claims brought forth by many of the supply companies regarding their products. One recent study by Middleton et al [32] evaluated the mechanical behavior of the orthodontic bracket, but focused on the area of the bracket/tooth interface. Few studies, have scrutinized the behavior and response of the orthodontic bracket in a mechanical sense. Questioning how does the bracket itself structurally respond to the torquing moment created by the twisting of the wire within the slot in both the elastic and plastic manners?

To achieve predictable clinical results, it is important for the orthodontist to know the magnitude of force moment generated by rotating a particular archwire within a particular bracket. Wire and bracket deformation will affect torque expression. In orthodontic treatment, the wire is twisted to engage the slot and torque expression is a function of the unloading characteristics as the wire returns to its neutral position relative to the bracket slot. Plastic deformation will result in reduced torque expression. Plastic deformation will effectively alter the bracket slot dimension which will further influence torque expression if the wire is removed and re-engaged. Bracket composition and bracket design is expected to influence the elastic and plastic deformation. To this date, there have been no published studies that have accurately reported orthodontic bracket deformation in relation to torque expression.

The first objective of this study was to develop a device capable of simultaneously recording torque expression and bracket slot dimension change (deformation). The second objective was to measure torque expression and bracket deformation for two identical bracket designs with different metal composition.

1.3 Research Questions

1 - Is there a significant difference in the magnitude of torque expression between two twin wing conventional ligation brackets of similar design, but of two different metal compositions: stainless steel and titanium?

2 – Is there a significant difference in bracket slot width change (elastic and plastic deformation) from application of a torsional couple via rotation of a wire within the bracket slot for identically designed stainless steel and titanium brackets?

1.4 Hypotheses

Null Hypothesis:

1. There is no difference in torque expression based on bracket metal composition
2. There is no difference in magnitude of slot width change (deformation) with torque application based on bracket metal composition.

Alternate Hypothesis:

1. There is a significant difference in torque expression based on bracket metal composition
2. There is a significant difference in magnitude of slot width change (deformation) with torque application based on bracket metal composition.

1.5 Review of Torque Expression and Deformation of Stainless Steel and Titanium Brackets

The majority of research regarding torque expression and potential bracket deformation from torsion of the wire rotating in the bracket slot has been directed at the investigation of the 'esthetic' orthodontic brackets. These 'esthetic' brackets encompass the polycarbonate and ceramic brackets, whereby a metal conventional bracket is often used as a control or gold standard. Alkire et al. [1] investigated the creep of brackets, finding that there was no significant difference regarding cumulative creep between the ceramic, the reinforced polycarbonate or the stainless steel bracket. Harzer et al. [7], also used steel brackets as a control when looking at slot deformation and equivalent torque capacities of polycarbonate brackets compared to a metal conventional bracket, finding that the metal bracket exhibited higher initial torque values. Other studies have found that the polycarbonate brackets can display upwards of 350% higher deformation than a conventional steel bracket at a moment of 15Nmm.[16] Feldner et al. [3] found that the stainless steel brackets do indeed demonstrate permanent deformation.

Focusing specifically on metal brackets, namely stainless steel and titanium, there are a number of published core articles, yet have been little recent advances. Kapur et al. [21] looked specifically load transmission and deformation of titanium and stainless steel brackets. Conventional brackets using elastomeric ties were tested with a 0.021" X 0.025" stainless steel wire at 15, 30 and 45 degrees regarding torque expression, using a custom apparatus. Initial and final bracket slot widths at the mesial end of the bracket were evaluated through the use of a travelling stereoscopic microscope. Only plastic deformation as a result of a 45 degree torquing rotation was examined, no analysis regarding elastic deformation was presented. He concluded that there were significant differences in bracket slot dimensions in both the stainless steel and titanium brackets, with the titanium brackets having less deformation. Furthermore, stating that only at the 45 degree mark did the stainless steel bracket express significantly more torque than the titanium bracket. The Ti brackets though expressed greater torque at the 15 and 30 degree intervals. He further presented that at 45 degrees the titanium brackets had begun to exhibit elastic deformation and it is due the rigidity of the stainless steel brackets that they exhibited the higher torque values.[21]

Several concerns arise from Kapur's conclusions.[21] No explanation was offered for the large difference in variance for stainless versus titanium brackets. The stainless steel exhibited standard deviations that were anywhere from 1.4 to 2.5 times greater than the titanium brackets. In addition, no data was accumulated throughout the experiment regarding the concurrent bracket slot dimensional change with torque expression and no quantification of elastic deformation was available. Furthermore, no analysis of the difference in initial slot width between the stainless steel and titanium brackets was offered. Looking at the presented data, it is clear that the titanium brackets had a mean initial slot width of 0.04 mm larger than the stainless steel brackets. As both brackets were tested with 0.0215 X 0.025 inch stainless steel wire, the initial increased width of titanium slot , creates play and subsequently a larger engagement angulation, which may in turn reduce torque expression and less stress within the titanium bracket structure than the stainless steel bracket at the same torsional angulation.

Prior to Kapur's study, Flores et al. performed a comparative study of the permanent deformation of metal brackets using a beam bending formula to calculate stress at failure.[21,22] Force required to permanently deform was determined as that point on the slope on the stress/strain graph began to decrease. The experiment used five different types of stainless steel, whereby he concluded that the 303S and the 17-4PH required significantly more force to deform. The material composition had a significant effect on force needed to permanently deform metal brackets. The conventional twin exhibited the greatest resistance to deformation compared to mini twin and modified twin designs, stating that the bulk and large size of the bracket allow forces to be dissipated through greater area resulting in less strain. The author concluded that the wing type design had a significant effect on the force needed for deformation.

Flores et al. [22] did not evaluate the elastic deformation process. Furthermore, similar Kapur's study, no mention is made as to the initial position of the wire within the slot.[21] No mention is made if the wire was perfectly centered or located closer to either the incisal or apical walls upon initial wire rotation. In addition, there was no investigation regarding archwire hardness versus that of the orthodontic bracket. This potentially is significant because if the wire is harder than the bracket, then notching of the bracket slot may occur, diminishing the amount of deformation that would have occurred if the wire and bracket hardness value were similar. Gioka et al.[5] presented that a low hardness wing component could reduce torque transfer from the archwire to the bracket. Furthermore, wear phenomena of a low hardness archwire or bracket slot wall, could prevent full engagement of the archwire with the slot walls and could give rise to permanent deformation of the tie wings.[5]

The relationship between torque expression and bracket deformation is a complex mosaic of many factors. It appears as though most published information regarding this is quite limited in scope, perhaps touching upon a few aspects, but in doing so, possibly excluding or neglecting others. There has yet to be a technique in which not only plastic but also elastic deformation of the orthodontic bracket slot can be quantified as a torsional force is being applied from a rectangular wire within the slot. Studies that have mentioned anything regarding elastic deformation in regards to metal conventional brackets have speculated in terms of amount and when it occurs. None as of yet have been able to capture a bracket's structural response to an applied torsional force and be able to record the resultant torque expression.

In order to provide concrete and logical conclusions regarding a bracket structure's response to a torquing moment it is vital to have a basic background knowledge regarding torque expression and deformation as they apply to the unique structural geometry and compositions of orthodontic brackets.

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Chapter 2 Review of Torque Expression

2.1 Torque

Defined as a twisting force that tends to cause rotation, the term torque encompasses many important aspects of orthodontic treatment. Control over the faciolingual positions of the dentition is pivotal for numerous reasons including post treatment stability, functional outcomes of treatment, peak esthetic result and the best health outcomes.[1] The term torque as utilized in orthodontics is used to describe the moment acting on a tooth to change the buccal-lingual root inclination.[2] Proffit presents that changes in root angulation in this plane are obtained through the torsion of an archwire in a bracket slot.[3] The torque moment is created through the twist of the archwire, termed torsion as depicted in figure 2.1. The amount of torsional force acting upon a tooth through the bracket/wire interface is termed as torsional load. The interaction between bracket and archwire develops a torquing moment, the degree to which is described as torque expression. Rarely, does the archwire fully obturate the bracket slot dimensions, leaving space between the archwire and the slot walls. The amount of space or angle present from a passive position in the bracket slot to that point at which both diagonal corners of an archwire engage the slot walls is often described as the engagement angle, deviation angle or play.[4]

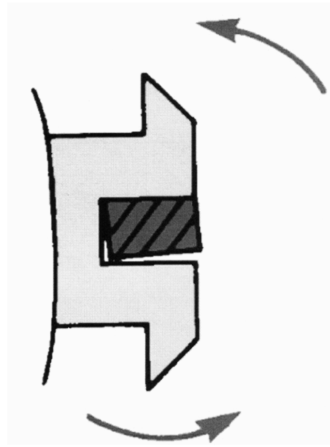

Fig. 2.1 Rotation of the archwire (torsion) within the bracket slot creates a moment of torque[3]

2.2 Measurement of Torque

A multitude of research methods and apparatus have been employed in an attempt to evaluate torque expression in an in vitro setting. At the present time, torque expression research is incorporating such technology as digital image correlation and finite elemental models to critically assess torque.[5,6] Initial attempts to examine torque such as that peformed by Steyn et al. [7] lacked the technology required to assess torque expression required with such small forces and angulations. Steyn's apparatus involved the use of pressure sensing devices on each tooth, and its calculation of wire angulation relied upon a mechanical setup with an indicator pointer.[7] Odegaard et al. noted in later work, that this early work by Steyn neither measured the engagement angle or the torque moment.[8]

Research that followed Steyn's early investigation continued to focus on engagement angulation between the bracket slot and the archwire. Using a torquemeter, Hixson et al. examined these parameters in three different brands of brackets using both new and recycled metal brackets.[9] Interest in the geometric cross section of the wires and its effect on both engagement angle and torque was at the forefront of research. Hixson et al. reported that the beveled edges of the archwires were responsible for the greater engagement angulations found is his work in comparison to the previously reported findings from both Dellinger et al. [10] and Creekmore et al. [11] who relied upon manufacturer specifications rather than measuring them. The effect of the wire cross sectional geometry, specifically the wire edge shape was further investigated by Sebanc et al. [12], whose findings supported that of Hixson regarding the effects of the wire bevel upon engagement angulation.

The first study to examine brackets with a prescription was performed by McKnight using an Instron machine.[13] The Instron machine evaluates mechanical properties of materials and components with applications to tensile strength, compressive strength, fatigue, flexural strength and impact force. The study methodology was based upon twisting the archwire within the bracket slot and reported that both increased wire torsion and wire dimension resulted in greater torque expression from all three bracket types tested.[13]

Inconsistentcy in the geometric dimensions of orthodontic archwires was reported by Meling et al. [14] Meling revealed the engagement angle was reliant upon the cross sectional dimension and edge bevel of the arch wire. Comparison between the measured engagement angles and the calculated ones based on specifications resulted in differences assigned to significant variation in wire cross sectional geometry.

Investigation into the effect of material properties of the archwire regarding torque expression was examined by Fisher Brandies et al.[15]

The study involved a multitude of wire dimensions from multiple manufacturers and focused on engagement angulation; reporting that measured engagement angles were up to 7.5° greater than their theoretical counterparts. Again wire size was reported as significant whereby an increase in archwire dimension led to a concurrent decrease in engagement angulation. Fisher Brandies et al. [15] reported that the wire edges were rounded and that the rectangular wire dimensions were on average undersized in both width and height.

The orthodontic measurement and simulation system (OMSS) was introduced by Drescher et al. in 1991, and is capable of measuring forces and moments in all three planes of space simultaneously from two force-moment sensors.[16] This system marked the introduction of computer controlled torque analysis in both static and dynamic fashions.

Both Gmyrek et al. [17] and Harzer et al. [18] have published studies using the OMSS focusing on both torque expression and bracket deformation. Their results have mimicked previous studies reporting that increased wire geometric dimensions led to greater torque expression. Much of their published research appears to be collaborative efforts to examine conventional polycarbonate and metal brackets. Their studies have revealed that polycarbonate brackets without metal reinforced slots do not provide adequate torque expression.

The OMSS has been and is still used in the study of torque expression. Increasing popularity of self ligation has led to studies such as Morina et al. [19] which examined two self ligating brackets, and three conventional brackets fabricated from stainless steel, ceramic and polycarbonate. Engagement angulation of the conventional metal bracket was reported to be 40% smaller than the other brackets in the study, yet no significant statistical difference in torque expression was noted between all bracket types tested. Bracket deformation in both the elastic and plastic states was not examined or accounted for.

Further exploration of torque expression of self-ligating brackets was performed by Badawi et al. [20] Construction of a novel device with near perfect alignment of the wire within the bracket slot in both the vertical and horizontal dimensions mated to a multi-axis force-torque transducer revealed a significant difference in engagement angle between active and passive self-ligating brackets. The study reported that the passive self-ligating brackets (Damon 2 and SmartClip) required up to two times greater wire twist angulation than the active self-ligating brackets (Speed and In-Ovation-R) to initiate torque expression.

Most recently Lacoursiere et al. unveiled the use of optical digital image correlation to examine bracket slot width deformation in both the elastic (temporary) and plastic (permanent) manners during the expression of torque.[5] Modifications including specifically developed software and hardware additions such as a stepper motor, a digital microscope and CCD image capture device. Lacoursiere et al. have proven that the use of optical digital image correlation is a viable technique in which to study bracket's response to a torquing moment through real time imaging.[5]

2.3 Determinants of Torque

Torque expression is dependent upon a multitude of factors. It results from a complex interaction of several determinants associated with the orthodontic archwire, the bracket and the dentition.

2.3.1 Archwire Effect on Torque Expression

Archwires formed of low modulus alloys including NiTi and β-Ti, are ineffective in efficiently transmitting torque moments through archwire contact points to bracket slot walls.[19] Stainless steel wires with greater torsional stiffness and increased hardness most effectively transmit torsion into the bracket.[21] Kusy (1983) through the use of nomograms showed that a 0.017 x 0.025 - inch stainless steel wire had seven times the stiffness of a similarly sized NiTi wire.[21]

Cross sectional geometry of the archwire, specifically the shape of the wire's corners influences torque expression.[13] An archwire may comprise of a specific width and height in cross section, yet the degree to which the corners are rounded and deviate from a true 90 degree square bend alters torque expression. As the corner of the wire increases in bevel, there is corresponding increase in torque play.[15] Subsequently resulting in reduced bracket wall engagement by a rotating wire, and concomitantly less force being transmitted to the bracket at a specific torsional angulation in comparison to a true 90 degree corner.[22]

Deformation of the archwire in either an elastic or plastic sense can lead to a resultant decrease in torque expression. Harzer (2004) described that such deformation gives rise to decreased force transmission into the bracket as a significant amount of the torsional energy is absorbed by the compliant material.[18] The transmitted torquing moment is consequently less reliably predicted with the use of a softer metallurgy rather than a hard variant in the archwire composition.

Lastly, wire dimension influences torque expression. Wire cross sections are often smaller than what is published by the manufacturer. Studies by Creekmore et al. (1993), Meling et al. (1997) and Odegaard (1994) all found that measured dimensions were markedly below manufacturer's specifications. [8,26,27] Such undersized archwire generates higher levels of torque play and brings about reduced torque expression.[12] McKnight et al (1994) further presented that larger dimensional archwires not only reduce the amount of torque play, but the larger wire results in a greater area of surface contact between the bracket slot walls and the archwire, yielding larger torque moments.[13]

2.3.2 Bracket Effect on Torque Expression

Sources of torque variation in the bracket include design, material properties of hardness and modulus of elasticity, manufacturing processes, and ligation method.[24]

Bracket design as it relates to geometric dimensions has been reported as affecting torque expression. McKnight et al. (1994) outlined that width of the bracket slot impacts torque moment generation.[13] A greater area of contact between wire and bracket results in more force transmitted through the bracket into tooth structure.[13]

Bracket composition significantly affects torque expression. Brackets fabricated of a composite, polycarbonate or polyurethane structure are considered not suitable for transmitting adequate torque as they often show deformation in the recommended torque range of 10-20 Nmm.[23] Flexibility of the bracket slot is likely the explanation and results in higher torque losses.[17,24] With regards to the straightwire technique, this category of bracket remains virtually ineffective with a torque play up to 13 degrees.[18] The incorporation of a metal slot within these types of brackets displays adequate stability to afford their use for torquing.[17]

Metal brackets of stainless steel and titanium alloy produce high initial torque moments and torque induced stresses are evident from internal slot wall notching and altered dimensions after torquing.[15,25] Brackets that are fabricated of low hardness metal alloy are susceptible to widening of the bracket slot under loading. Fischer-Brandies et al. (2000) expressed that their use represents a weak point especially when mated with archwires that can often be up to 3.7 times harder than the bracket, causing widening of the slot.[15]

Mode of ligation influences torque expression in both conventional and self-ligating bracket systems. Force relaxation of elastomeric ligatures has been patterned as having an initial exponential decrease that can approximate 40% loss in the first 24 hours after application.[28] Slack in the bracket - wire ligation results in potential incomplete and flexible wire engagement in the bracket slot further comprising torque expression.[24] Morina et al. in examining torque expression stated that a standardized manner was used to close stainless steel ligatures, whereby the ligature was adjusted so that the archwire was securely pressed onto the slot bottom and no play was obvious.[41] No mention though is made regarding any potential play in the closed ligature around the tie wings, nor is any mention made regarding potential deformation of the slot width with such ligation. Reduced slot clearance can be achieved through the use of steel ligatures throughout the range of slot-wire differences, precipitating greater torque expression.[15]

The design of the self ligation mechanism influences torque expression. Badawi et al. (2008), found that active clip designs appear to exhibit better torque control at clinically usable torsion angles in comparison to passive self ligation.[20] Engagement of the active clip upon the archwire in the slot reduces torque play. Furthermore, material stiffness of the clip influences the resilience of the clip to maintain archwire engagement in the slot. Clips fabricated of materials of greater elasticity, such as NiTi, may not exhibit reduced torquing moments when compared to stiffer materials such as stainless steel used in the clip. Huang et al (2009) reported that Speed brackets, incorporating a NiTi active clip did not express any appreciable torque until a torsional angulation of 15° to 18°, whereby the bracket began to exude behavior similar to other active self ligating brackets.[6]

Manufacturing processes have been reported to affect the accuracy of the specified torque values in brackets by to up 10%.[24] Considerable variation in manufacturer tolerances has been reported by multiple investigations, leading to reduced degree of torque control that would be expected or assumed as present. [27,29,30] Cash et al. (2004) reported that all brackets incorporated bracket slots that were anywhere from 5% to 17% greater in dimensions than reported by the manufacturer.[22] Brackets exhibiting actual dimensions that are larger lead to excessive and unexpected torque play, reducing torque expression and generating uncertainty in clinical practice, likely leading to decrease efficiency.

Furthermore, regarding manufacturing processes, it has been reported that presence of metal particles, grooves and striations are the result of the fabrication process.[24] Such irregularities present on the slot walls gives rise to a rough surface from imperfections which can preclude full engagement of the wire in the bracket and subsequently affect its dimensional accuracy. All of which can lead to inconsistent levels of torque expression, resulting in clinical inefficiency.[24]

2.3.3 Dentition Effect on Torque Expression

Interbracket distance has been shown to affect torquing moments. The interbracket distance is a function of tooth anatomy, arch characteristics and bracket geometry. Jarabak et al. (1972) reported on the effects of interbracket distance in relation to torque expression.[31] As distance between adjacent brackets increases, the torsional force developed by the wire decreases.[13] The greater the span, the larger the reduction in torsional force applied to the bracket slot walls.

Tooth morphology can influence torque expression as it relates to vertical bracket positioning on a tooth. Correct vertical placement of the bracket on a tooth (teeth) is critical to efficient success.[35] It has been reported that a shift of 1mm in the vertical bonding location of the bracket upon a tooth can affect the torsional angulation by upwards of 15 degrees.[32,33] Morrow (1978) in his thesis defended the idea that the morphology of dentition affects torque expression by examining the relationship of tooth morphology to the angle of the long axis between the root and crown of the tooth.[34] Large variations in tooth morphology exist throughout the population.[34]

Facial contour of the tooth influences torque expression in the straightwire appliance.[10] A bracket bonded nearer to the incisal aspect of a maxillary incisor, is generally on a flatter surface than what would be found nearer the gingival area. Such an incisal position would lead to higher initial torquing moments, as would be produced with a correct bracket location on a retroclined tooth.

2.4 Clinical Relevant Torque

There is a lack of definitive research regarding the minimum amount of torquing moment required to move a tooth effectively and efficiently. High levels of force have been reported as elucidating increased levels of pain, mobility and potential root resorption.[36] Gmyrek et al.have presented that a range from 5 to 20 Nmm is clinically effective.[17] The lower end of 5 Nmm directed towards incisor movement, while longer rooted teeth with greater surface area such as a cuspid needing a moment around 20 Nmm at the high end of the range.[34,37,38,39,40] It appears that the majority of published literature regarding torque expression has accepted this range as legitimate.

Research performed by Lee et al. demonstrated that there is a large degree of variability in the population's dental response to a torquing moment.[37] Such variability was related to a measurement of axial inclination induced from a specific amount of torque applied to the tooth over a period of time. Each orthodontic patient is a unique case entity entailing specific biological traits, functional and parafunctional habits, all of which may influence the amount of force required to move the tooth and the amount of time required.

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Chapter 3 Review of Bracket Deformation

The process of deforming an object whether it be in a temporary or permanent fashion is a complicated multifactorial event that is determined by factors such as material composition, magnitude and duration of force applied to the object, and geometric structure. In the field of orthodontics, the majority of discussion revolving around deformation is generally focused upon the archwire and its geometric response after having been engaged within the orthodontic bracket slots. Little information has been published regarding deformation of the orthodontic bracket.

3.1 Deformation

Deformation can be described as the action or process of changing in shape or distorting through the application of a force and is often described as strain. The applied force, or stress, can be in the form of tensile (pulling), compressive (pushing), bending, shearing or torsion (twisting). A typical stress-strain curve is depicted in figure 3.1.

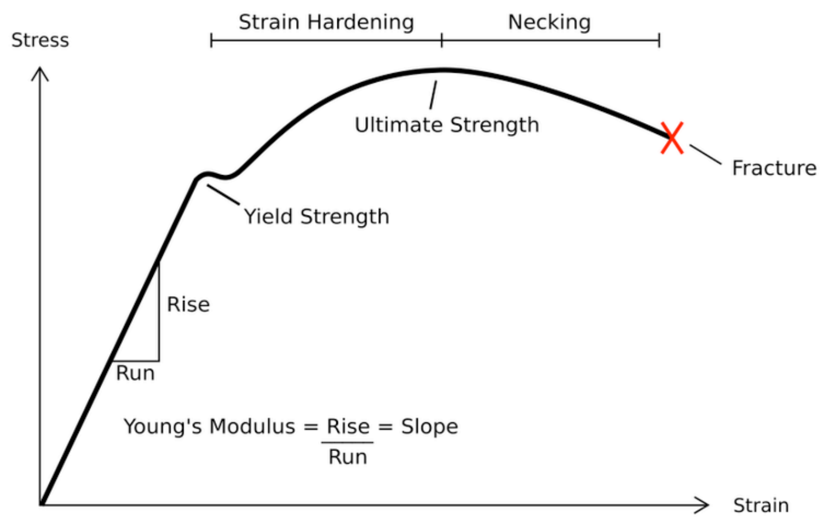

Figure 3.1 A Typical Tensile Stress - Strain Curve.
The elastic range depicted as the initial slope up to the yield strength, after which permanent deformation arises [31]

The application of a force or stress upon a body results in a proportional increase of inter-molecular forces arising within the body. Solids, such as metals are made up of crystalline lattice structure. Intermolecular bonds determine the material's modulus of elasticity. The counter reaction of the molecular structure to these forces forms a new equilibrium within the body structure. If the applied force is not greater than the maximum amount of counter force that can be created within the body, upon reduction or cessation of the applied force, the body will return to its original shape or state. This type of deformation is termed elastic as once the imposing force is ceased the object returns to its original form. The range of elasticity extends to the material's yield strength, the point at which plastic (permanent) deformation occurs as depicted in Figure 3.2.[46]

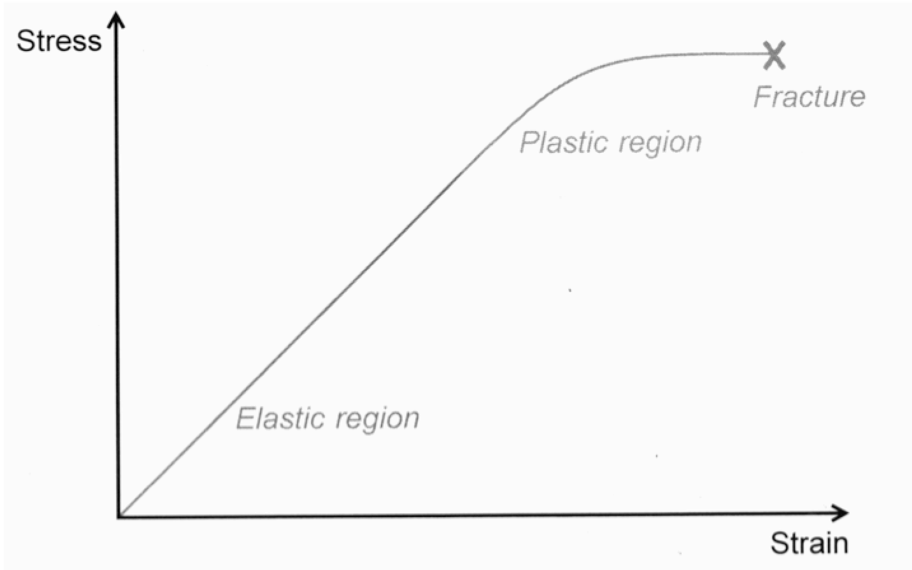

Figure 3.2 Stress-strain curve depicting the association between the force applied (stress) and the resultant deformation (strain)[31]

The object returning to its original form doesn't necessarily indicate that the structure has returned completely to its original configuration. Faults can arise within the structure. If the body undergoes numerous deformations, the stochastic phenomena known as metal fatigue can result. Metal fatigue as depicted in figure 3.3, describes the situation when after repeated deformations occur, which could be in the millions or trillions, small cracks result in the body. Although no obvious plastic deformation has occurred during this crack formation, the damage is cumulative and does not recover with rest. Fracture of the body is the common consequence. Factors such as material, shape, temperature, surface finish, applied stress range and how close each applied force comes to the elastic limit will influence the degree of metal fatiguing. [47]

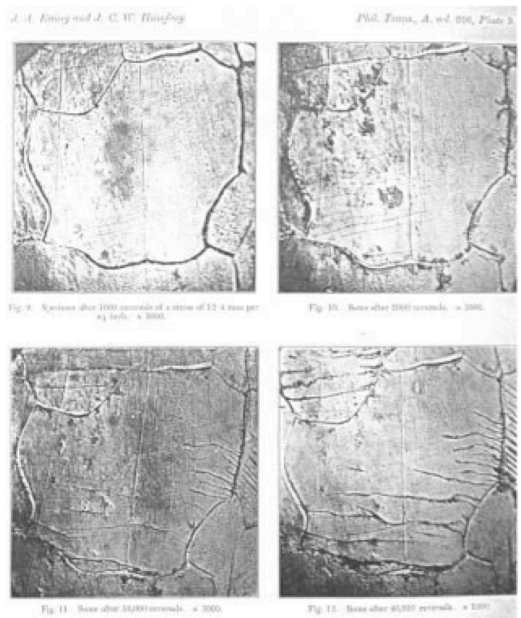

Figure 3.3 Metal fatigue cracks from repeated deformations [32]

If the stress exceeds the yield strength, then permanent or plastic deformation of the body will occur. [2,3] Generally a body that incurs plastic change will have undergone the full allowable elastic change prior to the plastic change taking effect, which is fully recoverable when unloaded. Plastic deformation involves two phases when created as a result of tensile stress. Strain hardening, the first phase, occurs as irregularities such a edge dislocations or screw dislocations occur within the crystal structure of a metal composed body, leading to alterations in the structural properties and often increasing their strength. The necking phase follows the strain hardening period and occurs as the structure's cross sectional area decreases at a faster proportion than the increase in strain within the material. This necking phase precedes the point at which the strained area either ruptures or it becomes so hardened from the strain that other points within the structure deform as a result.[48]

Another form of deformation, is known as creep. Creep is a time dependent phenomena, whereby the application of a stress that is below the yield point results in the accumulation of strain, subsequently creating the material to move slowly or deform permanently. The rate of this depends upon the material properties, the exposure time, temperature and the applied structural load.[49]

Creep in metals, generally requires an increase in temperature as well as prolonged time periods of applied force. Material creep is not a concern in this research as the testing was performed at room temperature. In addition, the duration of the applied torquing moment was very short with three seconds at each interval, resulting in less than two minutes of overall time per orthodontic bracket tested. These would be considered elevated temperatures or prolonged exposure time.

3.2 Orthodontic Bracket Design and Deformation

Orthodontic bracket design, both in material composition and geometric configuration varies according to manufacturer. All orthodontic bracket designs provide a slot for the archwire to engage the bracket and some mechanism to hold the wire in the slot. Interaction of the wire within the slot generates the necessary forces and moments to achieve desired tooth movements.

A conventional orthodontic bracket with its component parts is depicted in figure 3.4. The base or body of the bracket described as the large mass central component of the bracket, located between the archwire slot assembly and the bracket pad. The pad being the thin contoured portion that is bonded to the tooth structure, often having a mesh or textured surface to increase bond strength. The archwire slot is of rectangular shape, either being of 0.018” X 0.022“ or 0.022” X 0.028” dimension, is where the archwire slides into. The walls of the slot are formed from the base and the tie wing components. The tie wings extend in both the apical and incisal directions, providing a retentive means for either elastomeric or steel ligation.

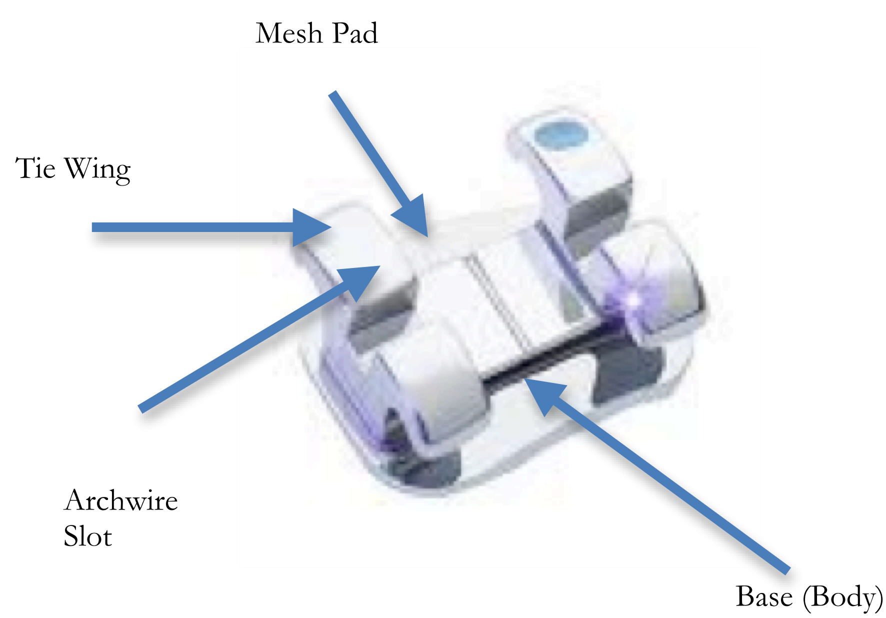

Figure. 3.4 Orthodontic bracket with list of components

Self ligating brackets appear to be gaining momentum with claims of increased efficiency and efficacy. [28,29,30] The conventional twin bracket design still appears to be the gold standard by which all others are judged and is not immune from design alteration with brackets appearing on the market with not four, but up to six tie wings. Brackets as depicted in figure 3.5 are available of different sizes regarding not only the facial (frontal) profile, but also the overall thickness of the bracket itself.

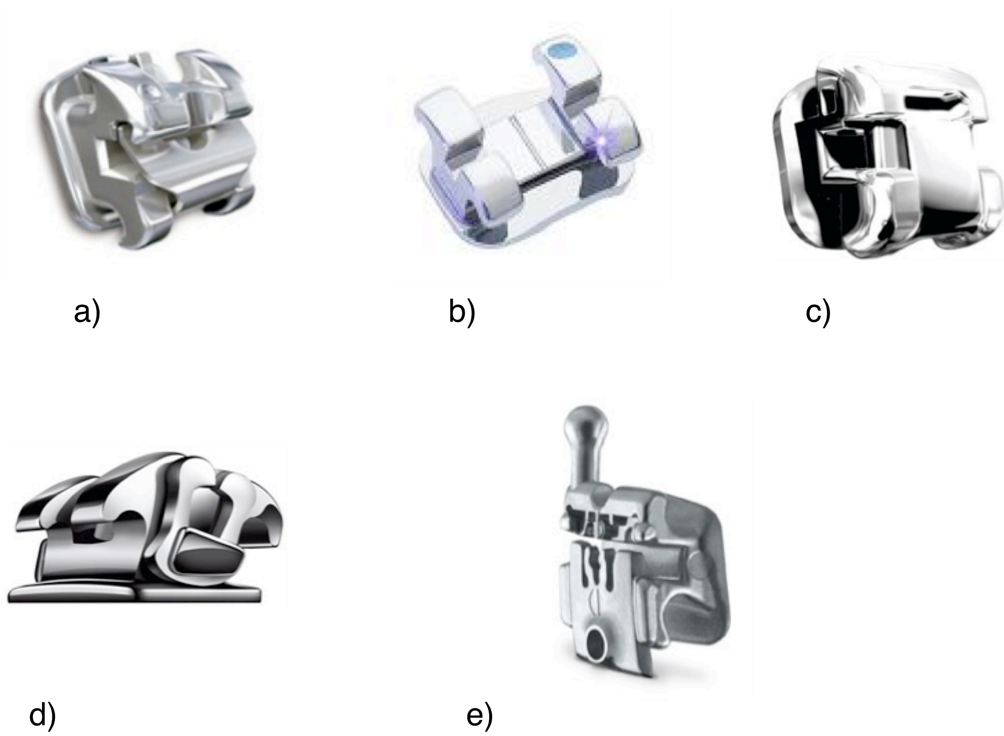

Figure. 3.5 Common bracket designs and profiles: a) Damon Q [33] b) GAC Innovation-R [34] c) Conventional Twin (OEM) [35] d) 3M Smartclip [36] e) Carrière SLB [37]

3.2.1 Bracket Design

Little published research is present regarding bracket deformation as it pertains to bracket design. Flores et al. [4] made generalized statements regarding bracket design pertaining to deformation, stating that the force needed to deform metal brackets was highest in the twin bracket design. Further, claiming that the reason for this observation was the large bracket size allowed the force to be dissipated through a greater area, minimizing stress. The larger size of the bracket does indeed aid in accommodating the induced stress from the archwire torsion, but more importantly the overall design of the bracket needs to be taken into account.

Examining the form of the bracket at the base-body zone, a larger body and attachment or junction with the bracket pad will resist bracket deformation about the x-axis, as would be induced from a torquing moment, more so than a narrower base or base-pad junction. Figure 3.6 depicts the axis geometry employed in this study. Whereby the x-axis runs from mesial to distal through the bracket slot. The y-axis is directed through the bracket from a facial to lingual aspect and the z-axis is found through the incisal to apical direction.

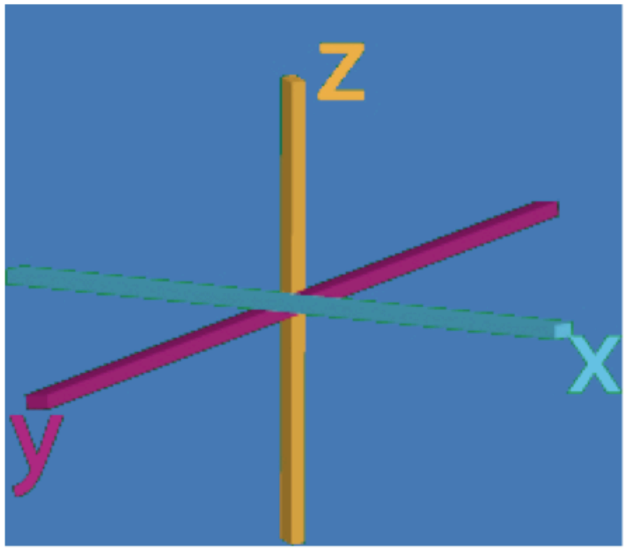

Figure 3.6 Dental depiction of the three axis (x,y,z) used in this study. The x-axis present from mesial to distal of the tooth, the y-axis from facial to lingual in direction and the z-axis present from incisal to apical.[50]

The same theory can be applied to the bracket tie wings or walls that form the bracket slot. Narrower or smaller dimensions of material are less likely to resist either elastic or plastic deformation from applied torsional force. This is further applied to the design of the bracket slot as in the conventional twin bracket there are four independent tie wings, and the only solid wall is that comprising the lingual aspect, which is adjacent to the mass of the bracket body. In such a design as the distance between the bracket pad/base and base of the bracket slot is minimal and quite broad in area, there will likely be little flexure at this point. Rather, the base of the tie wings becomes the fulcrum point and will exhibit the greatest amount of potential deformation from a torsional force due to the small mass of material present junction point between the base of the tie wing and bracket body which endures the majority of torsional force.

The structure of the bracket being made of a solid piece or a union of individual components could affect the bracket's ability to cope with induced stress. Potential clinical complications regarding bracket deformation or fracture may be linked to use of laser welding in titanium brackets where the tie wing body is welded to the base as research has shown the presence of pores and gaps at the junction site .[14] Such voids may create the potential for plaque accumulation, leading to increased potential for corrosion, whereas a solid one piece construction will likely provide greater strength and less potential for porosity. [14] This though would be dependent upon quality control of the chosen fabrication procedure. The majority of stainless steel brackets formed from the union of a body to base are done thru spot welding or by brazing with various solders such a 80-20 gold copper and 82-18 gold silver eutectics applied in either a wire or paste form.

3.2.2 Bracket Fabrication Method

The means of fabrication regarding the bracket may affect the bracket's resistance to deformation. The majority of brackets are either formed through a metal injection moulding process (MIM), machining or casting process.

MIM has been used successfully in the industrial field with the advantage of mass production of products of the same design, whereby precise near-net shape forming is required. [18,19,20,21,22] Numerous manufacturers employ MIM is bracket fabrication such as Ortho Organizers Delta Force brackets and Ormco Damon Q brackets.[33,37] The metal injection moulding process consists of feeding the specific formula of materials, often metal powders and plastic binders, into a heated barrel, where it is heated and mixed into a compound. The compound is granulated and then forced into a mold cavity where it is then cooled allowing the material to harden to a finished form, followed by debinding and sintering processes.

The products are up to 98% as dense as wrought metal and depending on the equipment and technology used, the as-sintered dimensional tolerances are +/- 0.3%.[13] Zinelis et al. (2005) presented that there is a high rate of variabililty in surface morphology amongst MIM brackets, which has significant effect upon bond strength.[42,43,44]

Further development is required to reduce the porosity found in titanium brackets on MIM fabrication, as it could cause higher standard deviations in hardness values, albeit at this time this porosity cannot be avoided in the powder metallurgy process displayed in figure 3.7.[23] Deguchi et al. [23] though have shown that a uniform distribution of hardness numbers can be achieved, thus suggesting that the sintering process can be uniform and isotropic.

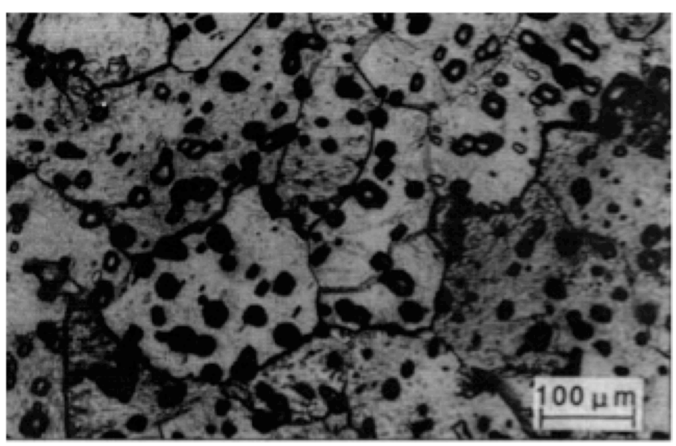

Figure. 3.7 Metallograph of sintered titanium powder showing porosities [31]

The machining process inherently involves the removal of product to create a final geometric structure. Such a process involves the use of such tools as lathes, milling machines and drill presses. Early ceramic brackets were created through diamond lathe machining processes, although current trends using sapphire based materials have resorted to injection molding procedures.[42] Techniques such as water cooling are used to minimize any adverse effects upon the metallurgical properties of the product being machined. Currently, new advances such as electronic discharge machining (EDM), uses a spark created between two electrodes to shape the material. The EDM process is often used for hard materials and alloys such as titanium.

The metal casting procedure involves pouring a liquid metal into an expandable or non-expandable mold. The mold contains a hollow cavity of the desired shape, wherein the metal is then allowed to solidify. The solidified part is the casting. This process is often used for fabricating complex shapes that would be difficult or uneconomical to fabricate by other methods. [24] Currently, most manufacturers employ this technique for molar tube fabrication, although G&H Wire Company fabricates their Zenith bracket with this process.[45]

3.2.3 Ligation Method

The method of ligation likely affects the bracket's ability to resist deformation specifically regarding the bracket slot width. A current search revealed no published research on the effect of ligature type on conventional bracket systems or self ligating design regarding bracket deformation. Most research involving ligation has been focused on force decay in elastomeric ligation and the effect of ligation on the expression of torque.[5]

In the conventional design, the most common means of ligation are the use of stainless steel ligatures and elastomeric modules or ties. The elastomeric tie, often of a polyurethane based composition, is flexible and loses much of its resilience or strength days after placement thus providing little resistance to bracket wall deformation or slot dimension change from the torsional forces of the wire. Moisture and heat affect the elastomeric module, exhibiting anywhere from 53 to 68% of force decay in 24 hours.[5] Some investigators have claimed that this force relaxation results in a reduction of force magnitude required to maintain a fixed strain.[6] In addition to this, the composition and initial dimensions of the module affect its general dimensional stability and resilience upon use.[5]

Stainless steel ligature available in different diameters, commonly used in either 0.009" or 0.01" diameter, could provide a degree of additional resistance to bracket slot deformation Binding the slot with wire may also redirect the torsional force to another location within the bracket if the wire is not fully seated. If such a transfer of stress occurs within the bracket, questions arise if the design of the bracket can handle the unexpected increased stress at a potential location within the geometry of the bracket that is not located at the base of the slot.

If there is potential to reduce the bracket slot width with the use of steel ligatures, this deformation in the slot could alter the prescription of the bracket.[8] To date there are no published reports regarding steel ligature plastic deformation of bracket tie wings. If such an event occurred, perhaps the repetitive ligation of the wire into the slot with stainless steel ligatures could affect the characteristics of the metal composition in the tie wings through work hardening of the base of the tie wings.

Analysis of the ligation design within the self ligating bracket has been reported recently regarding torque expression. The appropriate wire sequence is more important in torque expression than clip design used in the ligation mechanism.[9] The majority of self ligating brackets use either a NiTi alloy or stainless steel alloy for the clip mechanism. The clip design varies with particular manufacturer brands of brackets. The Damon MX (Figure 3.8) that utilizes a sliding steel door that positively locks into place creating a four walled lumen bracket slot. This design joins the incisal and apical walls through a strong positive clip union that may reinforce the rigidity of the slot, adding dimensional stability and strength against deformation of the slot from a torsional force created from rotation of the wire within the slot.

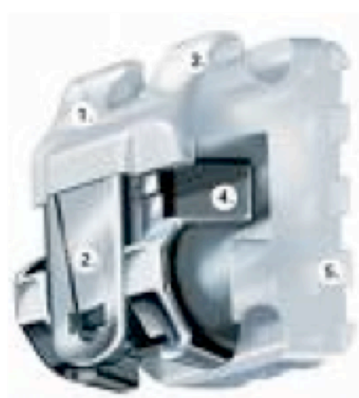

Figure. 3.8 Damon MX bracket showing sliding clip design [33]

Bracket designs such as the Speed bracket in Figure 3.9 a) (Strite Industries, Cambridge, Ontario) utilizes a sliding NiTi clip to ligate the wire into the slot. This clip design is similar to that of the GAC Innovation R (GAC, Bohemia, NY)(Figure 3.9 b) and Time brackets, whereby the clip positively locks into place upon closure but relies upon a small recess in the apical wall to prevent the clip from disengaging in a buccal fashion. This clip design likely does little to reinforce the bracket structurally to oppose bracket slot deformation from a torquing force as the orientation of the clip recess is in an apical-incisal direction, rather than a buccal-lingual plane that would oppose opening of the slot from torsion.

a) b)

Figure 3.9 a) Speed and b) GAC brackets showing clip design [40,34]

Other designs such as the Ortho Technology Lotus and Carriere LX, incorporate a sliding door mechanism similar to the Damon MX bracket, but have chosen to use a NiTi composition for the sliding door. These designs rely on the flexibility of the door design to slide past indentations in the facial aspect of the apical slot wall and create a positive lock. Any reinforcement of the rigidity of the bracket slot lumen from this design would be dependent upon the specific harness traits of the NiTi alloy, the amount of play and flexibility within the lock mechanism, and the depth and degree of the retention design.

3.3 Composition of Metal Bracket and Wire

Although several papers have been published regarding torque expression and deformation of aesthetic brackets of either ceramic or polycarbonate construction, a sparse amount of research has been reported regarding metal brackets and orthodontic wire composition.

The majority of metal brackets are fabricated of stainless steel or titanium. A number of stainless steel alloys including 316L, 2205 Duplex, PH17-4 and 304 types have been used for the manufacture of orthodontic brackets.[15,16,17] A study by Flores et al.[4] suggested that stainless steel orthodontic brackets are currently fabricated from five different American Iron and Steel Institute types of stainless steel.[10,11] The 300 series of stainless steels are the most common, whereby the 17/ 7 and 17/4 stainless steels are used as then can be heat treated. Parts made from machinable 300 grade stainless steel cannot be heat treated, thus limiting their strength in small cross sections and potentially limiting their application. Casting procedures using injection molding can use heat treatable 17/4 SS, and due to achievable exacting tolerances and strength, this process is highly favourable.

Similar to bracket design, there is a lack of published data on the effect of the metallurgical composition of either the orthodontic bracket or wire and their interaction regarding potential deformation as a consequence of applied torsional couples.[6] Flores et al. [4] have stated that the raw material composition of the bracket has the greatest effect on the force needed to permanently deform metal brackets. Their research design regarding the determination of permanent deformation load was conducted through a mathematical means based on the stress/strain curve, using the point at which the line's slope began to decrease.[4]

The majority of stainless steel brackets are manufactured from a two piece system consisting of a base and slot-wing component.[6] The base which includes the pad is often fabricated from a low hardness stainless steel with the theoretical application being that it simplifies bracket removal from dentition.[63] The tie wing-slot component is fabricated of a harder and stiffer stainless steel, such as a 17400 alloy to withstand wire engagement during torquing.[16]

Titanium brackets are also generally fabricated from a two piece process.. Zinelis et al. [64] has investigated the titanium bracket in regards to its compositional structure in two of the more common titanium brackets available, the Orthos 2 Ti and the Rematitan. Both brackets consisted of titanium within the base, but the Orthos 2, being a two piece laser welded bracket had aluminium and vanadium in addition to titanium in the wing-slot component.[64,65]

Gioka et al. examined a number of sources of variation of expression of torque in appliances, looking at such variables at slot dimension, the inability to fill the slot, wire alloy stiffness and ligation modes.[6] The examination of retrieved stainless steel bracket slots have demonstrated dimensional deformations and notched walls as a result of torque induced stress.[7,8] Recent published research by Huang et al.mentions that hardness and elastic modulus of the bracket are sources of torque variation and concludes that the torque angle/torque moment behavior is determined by characteristics of the archwire. No mention is made regarding structural integrity of either the bracket or archwire.[9]

The Vicker Hardness (VH) of the engaging archwire and that of the bracket may have serious implications regarding not only the potential deformation of the bracket, but also the potential torque expression from specific wire-bracket combinations. NiTi archwires generally have a Vicker's Hardness range from roughly 300 to 430 VH, and stainless steel wires can approach 600 VH. The Orthos Ti bracket wing-slot part approximates the VH of NiTi wire, whereas, the Rematitan has a VH much lower than either the NiTi or stainless steel wires. This may be significant regarding these brackets ability to withstand the strain created upon archwire engagement within the bracket slot, as the bracket slot hardness is less than that of the engaging archwire. Potential exists for bracket slot deformation, in the form of notching of the slot walls and or opening of the bracket slot.

The relationship of the hardness between the wire and bracket has been touched upon minimally by Fischer-Brandies et al. [31] where they concluded that a major reason for bracket slot widening with the application of a torquing moment was due to the difference in hardness between the wire and bracket. [8] Micro-hardness checks in the torqued brackets revealed an increase in the area of deformation. Increases of 5-10% in Vickers Hardness have been reported when testing used stainless steel brackets, being attributed to the cold work of the materials associated with their intraoral use.[63] Cold work hardening does entail a substantial degree of plastic deformation.[66] The plastic deformation observed led to amalgamation, which theoretically creates a state whereby increased force is necessary for deformation to increase with subsequent torquing force application.[67]

The action of rotating the orthodontic square or rectangular wire creates contact points between the edges of the wire and three bracket slot walls including the lingual, superior and inferior walls. Gioka states that there may be wear of the bracket slot and or the wire surfaces arising from the low hardness of alloys in the bracket could result in plastic deformation of wing. [6] Authors including Kapur-Wadhwa [7] have mentioned correct pairing of the orthodontic wire and bracket composition materials.[6]

With the correct pairing of materials used within each component according to hardness, one could diminish the possibility of either deforming the bracket or wire and consequently maximize the amount of force that reaches the periodontal ligament, rather than being lost due to strain. Manufacturers make no statement regarding the composition of their wires and brackets or correct mating of the two. For instance, if the stainless steel bracket is significantly harder than the wire that is engaging it, then the wire has a greater potential for deformation and subsequently, less torquing force is generated. In the opposite situation, whereby the wire is much stiffer than the bracket composition or design structure, leads to deformation of the bracket in either an elastic or, elasto-plastic fashion.[7] This being a direct consequence of the torsional moment generated from the engagement of the wire onto the bracket slot walls.

Material composition as it pertains to the bracket and wire hardness is an area requiring more research. Few conclusions have been drawn regarding the consequence of the modulus of bracket slot walls and torsional stiffness of the archwire, other than stating that it affects torque expression.[68] The relevance of this aspect of the straightwire appliance system and its consequences has not been researched adequately, leading to inadequate understanding and application. Further investigation is required in clarifying these interactions and how they relate to the width of the bracket slot, as this geometric variable will also have affect torsion expression and potential deformation of either the bracket and/or wire.

3.3.1 Stainless Steel Metallurgy

Stainless steel or inox steel is defined as a steel alloy containing a minimum of 10.5 or 11% chromium content by mass.[53,54] Stainless steel is available in over 150 grades and surfaces finishes specifically designed to suit the environment it is subjected to.[56] It does not rust or corrode as easily as ordinary steel and contrary to common belief, can stain. Carbon steel rusts upon exposure to air and moisture, due to a lesser percentage of chromium present. The larger amount of chromium present in stainless steel acts to form a passive film of chromium oxide that prevents further corrosion from occurring, by inhibiting corrosion from spreading into the internal structure. This protective film will quickly regenerate after the surface is scratched.

Austenitic stainless steels are the most common, comprising of over 70% of total stainless steel production. They are comprised of the 300 series and are typically an 18/10 composition, with 18% chromium and 10% nickel. Low carbon versions include the 316L and 304L and are used where biocompatibility is of concern.[55] The L represents that the alloy contains less than 0.03% carbon, reducing potential sensitization effect.

Martensitic stainless steels are stronger, but do not encompass the same degree of corrosion resistance. These are machineable and can heat treated to harden, but the process can lead to increased brittleness. The 17-4 PH stainless steel, as found in the Orthos bracket in the study, is a precipitation hardened stainless steel expressing extreme high strength and increased brittleness often used in orthodontic products.[6]

3.3.2 Titanium Metallurgy

Discovered in 1791 and named for the Titans of Greek mythology, it comprises 0.63% by mass of the earth's crust and is the seventh-most abundant metal.[60] Titanium (Ti) is an element with low density, corrosion resistance and the highest strength-to-weight ratio of any metal.[58,59] Often alloyed with other elements including iron, aluminium, vanadium and molybdenum, it has a range of applications spanning from aerospace to dental implants.[57]

ASTM International recognizes 31 grades of titanium, although there are more than 50 grades reported . Titanium alloy used in orthodontic materials is often Ti-6AL-4V (listed as ISO 5832-3)[42] due to its high tensile strength, high corrosion and crack resistance.[59,61]

Titanium is non-toxic and does not play any natural role within the human body.[62] It is non-ferromagnetic, allowing for the use of magnetic resonance imaging for the patient if required. Heat treatment is performed after the alloy is worked into its final shape to increase its final strength.

3.4 Clinical Relevance of Bracket Slot Deformation

Current orthodontic fixed straightwire appliances are precise instruments designed to provide precise function and efficiency for the clinician. There are currently a multitude of bracket prescriptions, such as Roth, MBT and Damon for example, from which to choose from. The aim of the prescription is to position each tooth within the arch with a specific tip, angulation and spatial orientation, so as to ideally employ a straight archwire and through execution of the prescription both the inter-arch and intra-arch dentition is ideally aligned. As brackets and their prescriptions are likely designed using population means regarding anatomical shape and position, rarely is it not necessary to incorporate either bracket repositioning and/or artistic detailing through bends in the archwire to achieve a satisfactory alignment of the dentition.

If clinicians are employing a specific bracket type and prescription with the conviction that each bracket will provide the advertised movement, they are relying on the assumption that the bracket and wire are both static entities. Published literature has presented the opposite, finding that brackets and wire deform in both a plastic and elastic manner.[4,6,50] These deformations could not only influence the attainment of a specific prescription, but also clinical efficiency regarding case completion.

Deformation of the bracket slot width during the expression of torque has been presented by authors such as Sadat-Khonsari et al. [69] and Harzer et al. [70] regarding polycarbonate brackets. They found that due to the amount of plastic deformation occurring within the bracket during the expression of clinically relevant torque, these brackets were unsuitable for transmitting the necessary torque.[69] Further emphasizing that due to the large degree of wire torsion required to achieve clinically applicable torque loads, that data should be provided by the manufacturer on the predicted amount of added or extra wire angulation required to achieve the expected level of torque at a particular wire angulation.[70]

Clinical relevance pertaining to deformation of metal brackets mimics that of the plastic brackets. Flores et al. [4] stated that although the clinician may be tempted to increase the slot torque to achieve quicker root movement, this increased load may deform the metal bracket. Such deformation whether it be in the elastic or plastic manner may cause opening of the bracket slot, consequently increasing torque play and reducing the torque effect.[71]

Elastic deformation of the bracket slot width during the expression of torque consequently results in lower than expected levels of torque expression leading to potential decreased efficiency.[71] Plastic deformation of the archwire slot dimension from excessive archwire torsion will give rise to an altered prescription, potentially reducing torque expression. An unexpected change in prescription can produce unpredicted tooth movement or lack thereof, both of which create decreased efficiency and effectiveness. If the clinician is unaware of either plastic or elastic changes occurring in the bracket from torque expression, there is no manner in which to compensate for this change to maximize the effects of the straightwire appliance with built in prescription as expected by the clinician.

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Chapter 4: Review of Digital Image Correlation

The majority of published research related to deformation of orthodontic materials utilizes SEM imaging techniques or traditional light microscopy. An alternative approach to image and track dimensional changes in orthodontic brackets is optical digital image correlation.

A digital image can be defined as a representation of a two-dimensional image using ones and zeros (binary). Without qualifications, the term "digital image" usually refers to raster images, which are more commonly referred to as bitmap images. [8] The 1960s brought forth the initial computer generated digital images, whereby they were used in space program development and medical research. [2] The introduction of microprocessors in the 1970s led to further advancement, eventually opening the avenue for the creation of image capture devices called charge-coupled devices (CCDs). CCDs are now widely used in an array of image capture devices such as dental radiography and photography. Currently, this form of imaging is now approximating the refinement of photorealism. [4]

4.1 Digital Image Correlation

Digital image correlation (DIC) was first conceived and subsequently developed at the university of South Carolina in the early 1980s. [5,6,7] DIC is an optical method to measure deformation on an object surface. It is a full field image analysis method that is based on a gray value pattern

located within predetermined zones, known as subsets or interrogation windows. The
subsets or windows are normally square and encompass a specific number of pixels both laterally (x) and vertically (y).[3]

This method tracks the gray value pattern (G (x,y)) in these subsets as the surface of the object undergoes deformation as in figure 4.1. The process as described begins with an initial image taken, the *reference* image, followed by subsequent images as the deformation process proceeds. It is predicated on the maximizing a correlation coefficient determined by examining pixel intensity within the subsets on two or more corresponding images: subsequently extracting the deformation mapping function that relates the images.

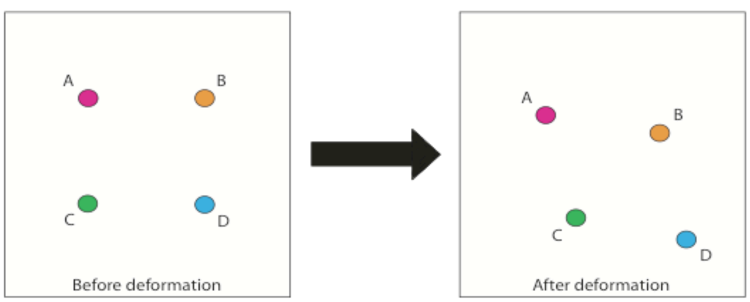

Figure 4.1 The four dots represent four different pixels. After deformation the same four dots are present, but now are other pixels. Software calculates and correlates the difference between the initial and deformed locations of the dots to determine the deformation.

To apply this method accurately, the specimen needs to be prepared by the application of a random dot pattern (speckle pattern) to its surface or the surface must be textured in a such a manner to exhibit a unique pattern, such as can be achieved through etching or air abrasion. Surfaces which exhibit high shine or gloss are difficult to pattern as pixel intensity may be of such a high value that although the subset has changed coordinates during the process of deformation, there is no change in the pixel intensity that can be monitored and subsequently correlated.

Simply stated, the process tracks and compares/contrasts the pattern and intensity of the unique dots in the reference image to the consecutive images taken during the deformation process (Figure 4.2).

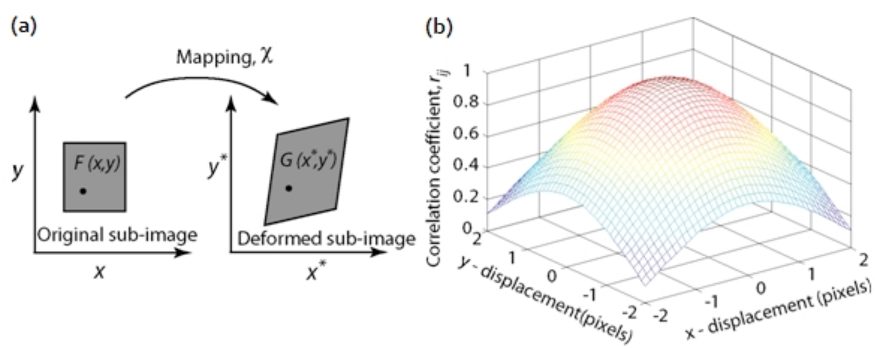

Figure 4.2 Image depicting deformation and a resultant change in coordinates, in conjunction with high correlation. [14]

4.1.1 Requirements for Optical Digital Image Correlation

The resolution that can be achieved through the use of an image based technique such as digital image correlation is dependent upon several factors. These include such things as a camera system with high resolution, high lens optical quality and an appropriate surface to be imaged/tracked which encompasses quality and appropriately sized marker traits.

High pixel count is a factor regarding image quality, but this can be misleading as pixel dimension influences resolution. Smaller pixel dimension results in better detection of higher frequency energy and provides greater image detail. The processing system that converts the raw data into the image is the most critical element as a higher quality system results will deliver greater accuracy. Other factors influencing image quality include the capture format (ie. RAW, JPEG), the capture medium (CMOS, CCD) and the quality of the lens (resolution, distortion, dispersion).

A pixel expresses a specific number of distinct colors, determined by the number of bits per pixel (bpp). In an image using simply 1 bit per pixel (1bpp), each pixel can either be on or off, thus representing 2 colors or shades. The addition of one bit, results in doubling of the number of colors or shades available. Therefore an image with 2 bpp could express 4 different shades or colors and subsequently, the 8 bpp express up to a maximum of 256 shades. Consequently, a higher bit rate produces significantly greater image accuracy at an exponential rate.

4.1.2 Cross Correlation Coefficient

The process of cross correlation is simply that of comparing subsets of numbers between two digital images and measuring how well the subsets match. The numbers within each subset represent the value of intensity given to each pixel within the interrogation window. [6] The coordinates or grid points (x_i ,y_j) and (x_i^* ,y_j^*) are related by the deformation that occurs between the two images as in figure 4.2. The operation of a CCD camera results in discrete information being obtained, thus no information regarding gray level between pixels is acquired and entails the use of sub-pixel interpolation to smooth out these areas.

4.1.3 Digital Image Correlation Process

The process is as follows:

1) A digital image is captured encompassing specific pixel dimensions
2) Gridwork of predetermined size is applied to the image, creating subsets or windows of specific pixel quantity
3) The grayscale pattern within each subset is recorded at specific time intervals, such as T_0, T_1....Tn
4) These subsets are then cross correlated, creating a correlation map for each interrogation window
5) A peak detection algorithm is used to determine the maximum correlation peak within the data by comparing the luminosity distributions between the subsets

6) Displacement vectors are determined by analyzing the subsequent distance and direction form the center of correlation

7) The displacement vectors then form a map of displacement for each subset

8) Displacement vectors can then be converted into a stress measurement and a scalar map of stress components

4.2 Applications of Digital Image Correlation

The current pace of technological advancement in research is demanding for higher complexity to be encompassed in even smaller space requirements in product development. This drive towards miniturization down to the nanometer dimensions has created the challenge of measuring strains in such small devices. The use of conventional extensometers and resistance foil gages could be damaging or not possible for such small scales.

DIC technique expresses the following features is extremely desirable:[8,]

- no contact with the specimen required [10, 11]
- sufficient spatial resolution to measure locally at the region of interest
- the ability to capture non-uniform full-field deformations
- a direct measurement that does not require recourse to a numerical or analytical model.
- ever increasing quality and decreasing cost using CCD to CMOS sensor based cameras [10]
- images with a high frequency [12,13]

DIC offers characterization of material parameters far into the range of plastic deformation. Its powerful data analysis tools allow the determination of location and amplitude of maximum strain, which are important functions in material testing. DIC is also ideal for fracture mechanics investigation and full field measurement delivers exact information about local and global strain distribution, crack growth and can be used for fracture mechanics parameters

4.3 Accuracy of Digital Image Correlation

DIC is currently incorporated in a vast array of fields of study for future research providing proof of the accuracy and validity of this technique. Roux et al. states that displacement uncertainties of the order of 10^{-2} pixel are achieved.[9] This allows the optical image correlation technique to reveal cracks that otherwise cannot be seen if their opening is less than one pixel wide.[9] Subpixel accuracy is achievable, with accuracy of 0.05 pixel was shown by early developers of this system using an 8-bit digitizer.[1,6]

Digital noise, can be minimized or removed through the use of multiple initial reference images and analyzing the raw differences, defined as the time average method. [1] Relying on analytical or numerical elastic modeling is extremely beneficial in lowering uncertainties and the detrimental effect of noise inherent to any measurement technique. It has been stated that the accuracy of the system could reach 0.1 to 0.2 image element, instead of 0.01 image element due to noise.[1]

The reliability of the correlation measurement has been validated through studies such as that done by Yang et al. Their research found that the accuracy was 0.1 *u*m, and when using pixel size to validate, the accuracy of the system was a miniscule 0.21% error.[1] The digital image correlation technique is accurate and its reliability is supported by a vast array of applications in the manufacturing sector.

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Chapter 5 Torque Expression and Slot Width Deformation of the Orthos and Orthos Ti Bracket Systems

5.1 Introduction

In orthodontic literature, torque is often defined as either the buccalpalatal root inclination of the tooth or the force generated from twisting of a rectangular archwire in a rectangular bracket slot.[1,2] However, torque is more accurately defined as a moment or moment of force, or the tendency of a force to rotate an object about an axis.[3]

Correct tooth axial inclination in the buccal/lingual plane, is important for both the anterior and posterior dentition.[4] Anterior dentition, primarily that of the maxillary arch is greatly affected in both an esthetic and functional means by appropriately torqued incisors.

The clinically appropriate range of torque expression has not been well defined, but the majority of the published research states that a range spanning from 5 to 20 Nmm is needed to efficiently torque an anterior incisor. [5,6,7,8,9,10] Torque expression has been reported to be related to bracket slot and wire dimensions, wire and bracket properties, inter bracket distance and the relative twist of the wire relative to the bracket slot.[11,12,13,14,15,16,17,18]

Deformation is defined as the action or process of changing in shape or distorting through the application of pressure and can be either temporary or permanent. A permanent deformation is described as irreversible change in shape in response to the applied force, and it termed as plastic deformation. Temporary or elastic deformation, is a change that returns to the original shape once the applied force causing the deformation is removed.

To date, published literature reporting torque expression and bracket deformation has focused primarily on esthetic brackets.[2,5,19,20] Limited studies are present which examine these characteristics in metal brackets.[21,22,23] The measurement of deformation in bracket slot width during either the loading or unloading phase of an applied torquing force has been presented by Lacoursiere et al. (2010) through the use of digital image correlation.[23] Prior to this study, most investigation into bracket slot changes were capable of only measuring plastic change, and were unable to evaluate the overall deformation throughout the loading and unloading ranges. [4,21,22,37]

The use of titanium metallurgy in the fabrication of orthodontic brackets has become more commonplace.[24] Reasons for its use include such traits as greater corrosion resistance, absence of allerginicity, a high rate of biocompatibility and perhaps marketibility.[25,26,27] Titanium has proven to be a valuable material for biomedical applications with a variety of uses in dental implants, fixation screws and plates used in orthognathic surgery and even arthroplasty materials.[28]

The majority of stainless steel orthodontic brackets were produced using Type 304 AISI (American Iron and Steel Institute), due to many appealing qualities which include superior resistance to fracture and deformation.[29,30,31] The current trend is to fabricate steel brackets using a steel such as 304 or 316 AISI in the base of the bracket, studies have supported the theory that such metals expedite debonding the bracket, .[32,32] The wing component is formed from a stiffer material such as

17400 alloy, Eliades et al. (2002) stating that such metal will withstand clinical loads during wire activation.[34]

The aim of this study was to evaluate the loading and unloading torque expression and bracket slot deformation characteristics of the Orthos orthodontic bracket in both the stainless steel and titanium versions. Specifically examining the following:

1) The overall torque expression patterns of both the stainless steel and titanium brackets.
2) The difference in torque expression between the stainless steel and the titanium bracket at specific collection point angulations.
3) The difference in bracket slot width change between the stainless steel and titanium brackets at specific collection point angulations.
4) The overall and plastic (permanent) bracket slot width changes in both the stainless steel and titanium brackets.

The overall bracket slot width change is defined as the maximal change in bracket slot width dimension occurring throughout one complete test cycle. The plastic (permanent) bracket slot width change is defined as the difference in bracket slot width between the dimension determined at a wire torsional angulation of 0° at the start of the test cycle and the dimension measured at 0°rtn at completion of one complete test cycle.

5.2 Materials and Methods

The study sample consisted of sixty (60) first right maxillary incisor brackets in 0.022” X 0.028” slot dimension, 15° torque and 5° tip prescription, of which 30 were Orthos and 30 were OrthosTi brackets (Ormco, Orange Calif).

The moment of force was created by twisting a 0.019” X 0.025” stainless steel archwire (Ormco, Glendora, CA) held in the bracket slot with elastomeric ligation. Each bracket / wire combination was tested throughout a wire angulation range of 0° to a maximum of 51° (loading phase) and then returned back to 0° (unloading phase) at 3° intervals resulting in 36 data points. Moments and forces in all three planes of space were recorded at each data point. Bracket / wire combinations were tested in a randomized order. Final sample size consisted of twenty eight (28) stainless steel and twenty nine (29) titanium brackets. The elimination of three brackets was due to fracture of the epoxy resin during the testing procedure.

Testing was performed using the torque measurement device previously described by Lacoursiere et al.[23] and the experimental setup is shown in Figure 5.1. Initial neutral positions were kept near zero as allowed by the inherent limitations of the system. Use of the live feed CCD camera imaging of the bracket from an overhead perspective enhanced neutral alignment of the orthodontic wire within the slot.

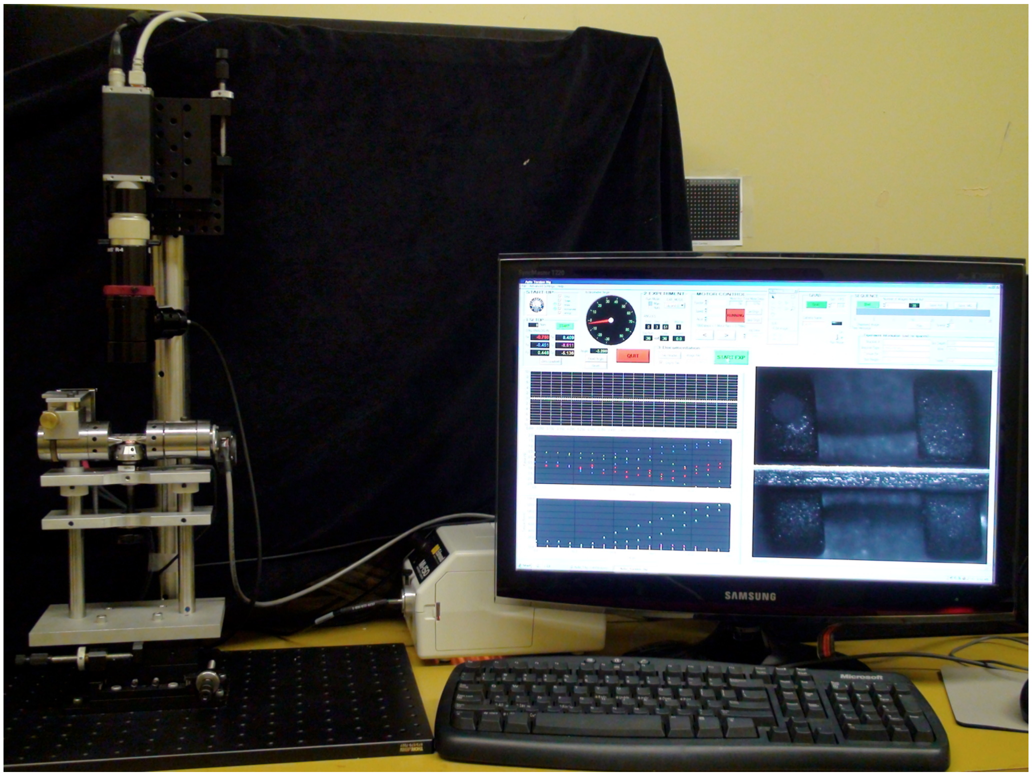

Figure 5.1 Modified orthodontic torque measurement device and custom software

Offset of the load cell compared to the position of the bracket was accounted for in all three planes of space through a mathematical transformation applied to the torque expression data. The transformed data was recorded in a spreadsheet (Microsoft Office Excel 2003, Microsoft Corp. Redmond, WA, USA).

Deformation analysis relies upon the process of digital image correlation to determine the relative change in the bracket surface location as it relates to movement of the bracket slot. Digital images were processed using commercial software (LaVision GmbH,DaVis 7.2, Göttingen, Germany,2007) through algorithms that cross-correlate contrasts between images to create displacement vector files.[35,36]

A custom code (Matlab ,The Mathworks Inc. , Natick, MA, USA) was used to determine bulk movement of tie wings . This data was output to a spreadsheet for further processing and data display. An analysis region is chosen on each tie wing, which delineates the area analyzed regarding deformation of the bracket slot. Figure 5.2 shows examples of typical digital images of the bracket/wire combination and the associated analysis regions. Tracking of the analysis regions identifies any deformation of the slot from torque expression throughout the test range. The distance between the right hand side regions is averaged with that of the left hand regions providing a linear measurement of overall bracket slot deformation.

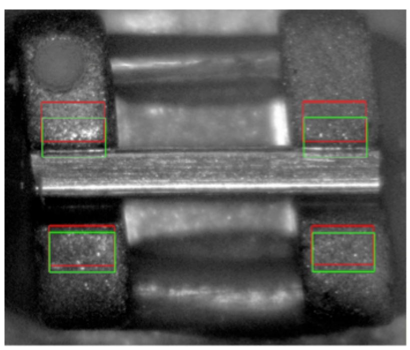

Figure 5.2 Graphics user interface displaying analysis box regions, colors representing particular points of interest: Green: Initial neutral position at 0° wire angle Red: Maximal deformation or bracket slot width

A statistical package (SPSS; version 18, Chicago, Ill) was employed for data analysis. The model of repeated measures ANOVA was employed to analyze the slot deformation and torque expression data. A significance level of 0.05 was used, P-values greater than 0.05 were considered to be statistically non significant.

Data analysis focused primarily on nine collection points consisting of wire angulations of 12°, 24°, 36°, 48°, 51°. Collection point angulations after reaching the maximum angulation of 51° notated with rtn, which stands for return to neutral.

5.3 Results

Torque Expression

To answer the objectives we applied repeated measures ANOVA. Normality assumption of the torque expression was reasonable based on boxplot as well as Komolgorov-Smirnov tests. However equal variance assumption was not met for some collection point angulations, and caution is used regarding interpretation of the statistical result.

Descriptive statistics regarding both brackets for the nine collection points used for statistical analysis are reported in Table 5.1. Appreciable variations were found for both bracket compositions with the steel bracket standard deviations ranging from 1.41 to 7.48 Nmm. Titanium brackets expressed greater variation with standard deviations ranging from 4.86 to 12.99 Nmm. The greatest variations occurred at a wire rotation of 36° for titanium bracket and the steel at 24°.

Table 5.1 Torque expression descriptive statistics of the Orthos (SS) and OrthosTi (Ti) brackets with sample sizes of n = 28 and n= 29 respectively.

Angle	Stainless Steel (SS)	Titanium (Ti)
	Mean (Nmm) (SD)	Mean (Nmm) (SD)
12°	4.085 (3.079)	5.663 (6.474)
24°	30.094 (7.478)	31.267 (9.606)
36°	65.223 (7.435)	68.987 (12.989)
48°	89.664 (4.488)	95.237 (9.859)
51°	92.057 (5.062)	98.994 (10.086)
48°rtn	77.452 (4.922)	84.339 (9.925)
36°rtn	30.047 (4.316)	35.789 (8.873)
24°rtn	5.515 (2.347)	10.479 (7.536)
12°rtn	2.241 (1.409)	4.281 (4.860)

The repeated measures ANOVA shows that with a P value of < 0.0001 the wire angulation is a significant factor in torque expression. Interaction between wire angulation and metal type is also significant with a p value of .017. In addition, metal composition of the bracket is also significant (p value 0.012) regarding overall torque expression.

Multivariate tests regarding angulation and metal type found that the torsional angulation of the archwire accounts for up to 98.7% of the variance in the torque expression data sample. The interaction between wire angulation and metal type accounts for up to 6.3 % of the displayed variance. Metal composition of the bracket accounts for 11.0% of variation in torque expession.

Examination of the bracket torque plots (Figure 5.3), displays that both brackets exhibit similar torque expression curves. The Orthos Ti titanium bracket produced greater torque values than that of the steel variant throughout the test cycle, with a maximum of 6.937 Nmm difference at 51°. Torque expression values were significantly different from 48° to 24°rtn data points. (Table 5.2)

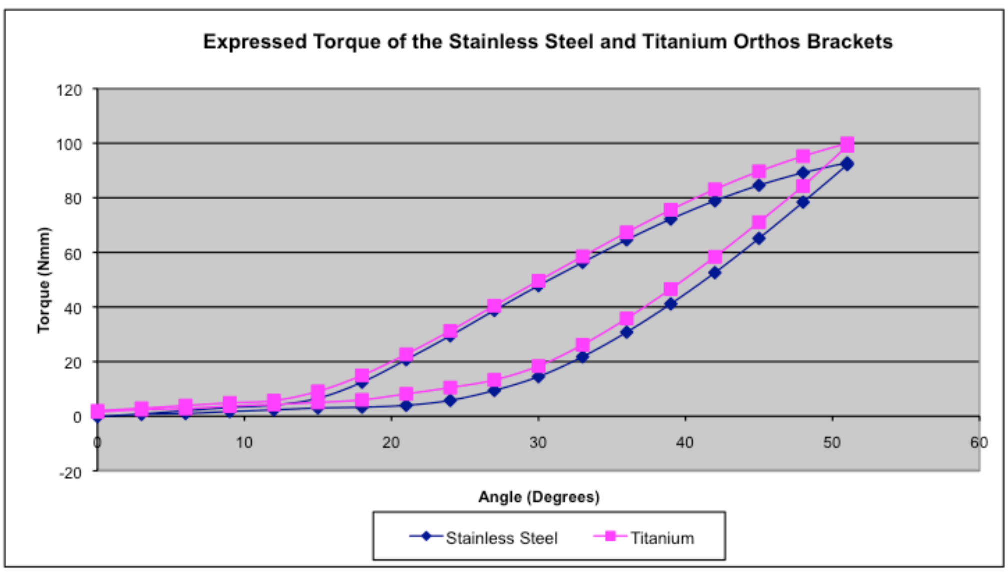

Figure 5.3 Expressed torque plots of the stainless steel and titanium Orthos brackets

Table 5.2 Comparison of torque expression between titanium (Ti) and stainless steel (SS) metallurgy at each collection point angulation

Angle °	**Mean Difference**	**P Value**	**95% CI for Mean**
	(Ti – SS) Nmm		
12°	**1.578**	**0.248**	**(-1.129 , 4.235)**
24°	**1.173**	**0.610**	**(-3.408 , 5.754)**
36°	**3.765**	**0.187**	**(-1.881 , 9.411)**
48°	**5.573**	**0.008**	**(-1.482 , 9.664)**
51°	**6.937**	**0.002**	**(2.677 , 11.197)**
48°rtn	**6.887**	**0.002**	**(2.705 , 11.069)**
36°rtn	**5.742**	**0.003**	**(2.016 , 9.467)**
24°rtn	**4.964**	**0.002**	**(1.978 , 7.949)**
12°rtn	**2.040**	**0.114**	**(-0.504 , 4.585)**

Bracket Slot Width Deformation

As the experimental torsional angle increased, so did the amount of variance present in each sample. Bracket #42 of titanium structure appeared to be a potential outlier. Statistical analysis was performed to ensure that its incorporation into the sample did not significantly affect any outcomes. No significant changes were noted with the inclusion of bracket #42, therefore it remained within the sample.

The model of repeated measures ANOVA was executed to investigate the data sample. Both Levene's test of equality of error variance and residual plot analysis confirmed an equality of variance and uniform data distribution.

Table 5.3 presents the descriptive statistics for each bracket type. In both bracket types a general trend exhibited an increased variation coincident with increased wire angulation. Titanium brackets having a standard deviations ranging from 0.004 to 0.024 mm and stainless steel brackets ranging from 0.001 to 0.024 mm. Mean quantification of slot width change resulted in a range of 0.001 to 0.077 mm for the steel brackets and 0.0003 to 0.068 mm in titanium. Steel brackets reaching a maximum slot change at 51° and titanium at 48°.

Table 5.3 Bracket slot width change descriptive statistics of the Orthos (SS) and the OrthosTi (Ti) with sample sizes of n = 28 and n = 29 respectively

Angle	Stainless Steel (SS)	Titanium (Ti)
	Mean (mm) (SD)	Mean (mm) (SD)
12°	0.001 (0.001)	0.000 (0.004)
24°	0.016 (0.004)	0.022 (0.005)
36°	0.041 (0.007)	0.044 (0.014)
48°	0.077 (0.024)	0.068 (0.019)
51°	0.077 (0.013)	0.068 (0.024)
48°rtn	0.075 (0.013)	0.065 (0.023)
36°rtn	0.062 (0.013)	0.052 (0.024)
24°rtn	0.040 (0.014)	0.022 (0.016)
12°rtn	0.038 (0.016)	0.011 (0.009)

Figure 5.4 depicts the loading and unloading graphs relating to the slot width change in both brackets. The stainless steel brackets exhibit less slot deformation throughout the mid range of the loading curve, but surpass the titanium at the highest torsional angulations. It is evident that the steel brackets express a greater degree of permanent slot change throughout the unloading phase. The titanium brackets present a unique unloading curve. In the midrange of the unloading phase from collection points 45° to 24° a small amount of slot width change is evident, but from 24° to 18° the mean width becomes less than that in the loading phase. Past 18°, the unloading slot widths surpass those of the loading phase displaying the permanent change resulting from the archwire torsion.

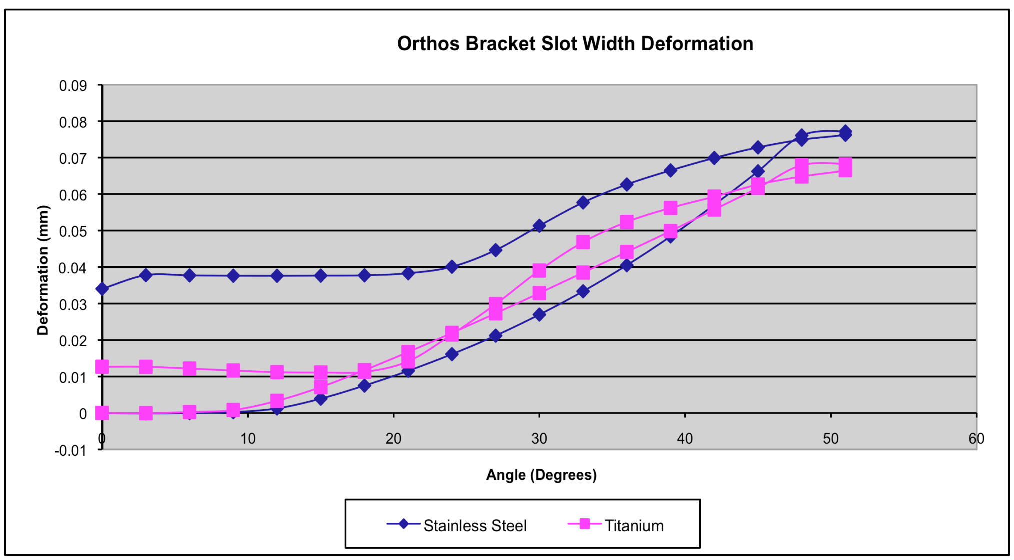

Figure 5.4 Bracket slot width deformation plots of the Stainless Steel and Titanium Orthos brackets

Mean difference in slot width change between steel and titanium brackets was found to be significant at the 12° and 24° collection points during the loading and unloading stages of a test cycle as seen in Table 5.4. Titanium brackets presented greater slot deformation during the loading stage up to 36°, at which point the steel brackets exhibited greater slot width change throughout the remainder of the test cycle back to the initial neutral position.

The repeated measures ANOVA shows that with a p value of < 0.0001 the wire angulation is a significant factor regarding slot width deformation. Interaction between wire angulation and metal type is also significant with a p value < 0.0001. Metal composition of the bracket is also significant (p value 0.020) regarding bracket slot deformation.

Multivariate tests regarding angulation and metal type found that the torsional angulation of the archwire accounts for up to 90.7% of the variance in the data sample. The interaction between wire angulation and metal type accounts for up to 28.0 % of the displayed variance. Metal composition of the bracket accounts for 9.5% of variation of the deformation.

Table 5.4: Comparison of bracket slot width change (mm) between stainless steel (SS) and titanium (Ti) metallurgy at each collection point angulation

Angle °	**Mean Difference**	**P Value**
	(SS - Ti) mm	
12°	**- 0.002**	**0.024**
24°	**- 0.006**	**0.000**
36°	**- 0.003**	**0.274**
48°	**0.009**	**0.088**
51°	**0.009**	**0.080**
48°rtn	**0.010**	**0.050**
36°rtn	**0.010**	**0.056**
24°rtn	**0.018**	**0.000**
12°rtn	**0.026**	**0.000**

Overall and plastic (permanent) bracket slot width change is presented in table 5.5 and table 5.6. Stainless steel brackets expressed significantly more plastic deformation with a mean slot width change of 0.038 mm, nearing three times that of the titanium brackets. Steel brackets corresponding standard deviation was also more than twice that of the titanium regarding plastic deformation. Steel brackets displayed greater total deformation change but was not a significantly difference with a corresponding p value of 0.091.

Table 5.5: Overall and plastic bracket slot width changes descriptive statistics for the Orthos (SS) and OrthosTi (Ti) brackets with sample sizes of n = 28 and n = 29 respectively

Deformation Type	**Stainless Steel (SS)**	**Titanium (Ti)**
	Mean (mm) (SD)	**Mean (mm) (SD)**
Plastic (permanent)	**0.038 (0.016)**	**0.013 (0.007)**
Overall	**0.077 (0.013)**	**0.068 (0.012)**

Table 5.6: Comparison of plastic and overall bracket slot width change between stainless steel and titanium brackets

Deformation Type	**Mean Difference**	**P Value**
	(SS – Ti) mm	
Plastic (permanent)	**0.025**	**0.010**
Overall	**0.006**	**0.091**

5.4 Discussion

Examination of the Orthos conventional bracket design which entails flat surfaces on the facial aspect of the tie wings allows for easier focusing on the whole bracket versus a contoured bracket such as a GAC Innovation R for example. This results in increased accuracy regarding image processing and subsequent vector generation. Similar standard deviations in overall slot width change between both the steel and titanium versions were recorded. Higher variability regarding plastic deformation was found and could attributed to greater unpredictability between individual tests. Data collection points being measured at intervals of 3° within the experiment confines any conclusions drawn regarding torsional angulations to be estimated to within 3°.

Orthos Stainless Steel

Figure 5.5 depicts the torque expression plot of the stainless steel brackets. Between 12° and 15° of wire angulation the steel brackets express the minimum amount of torque required to move a tooth. In this range the archwire engages the slot walls and adequate torque levels are generated with a small degree of wire torsion. The loading curve remains flat until 12° when torque expression begins a linear ascent to around 45°. Throughout this range, the corners of the archwire have fully engaged the bracket slot walls, producing the linear increased in expressed torque with increasing torsional angulation. The torque curve appears to commence flattening after 45°, which may be the result of bracket slot and/or wire deformation. Another potential explanation may be that the archwire's diagonal reference is nearing perpendicular to the base slot wall and if it were to pass through this perpendicular point, torque expression may correspondingly begin to decrease.

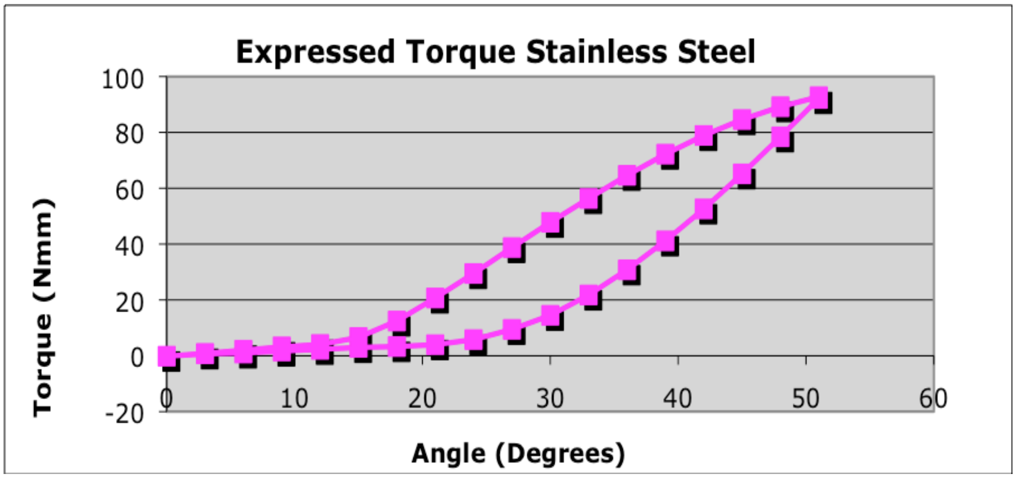

Figure 5.5 Plot of Orthos stainless steel torque expression (Nmm)

Upon rotating, once the archwire has engaged the slot walls at 15° the steel brackets show increases approaching 8 to 10 Nmm for every additional 3° of wire rotation throughout the linear section of torque expression. Minor increases in wire rotation created large increases in torque, which could lead to potentially high levels of pressure and discomfort for a patient.

Although not apparent to the naked eye without magnification, examination of figure 5.6 which displays the magnified images of a steel bracket at every second collection point, the bracket slot deformation (widening) is clearly evident in the higher wire angulations. Figure 5.7, depicts the bracket slot deformation plot for the steel brackets and displays a linear trend that commences at 15° and continues to the 48° mark. This curve closely approximates that of its torque expression.

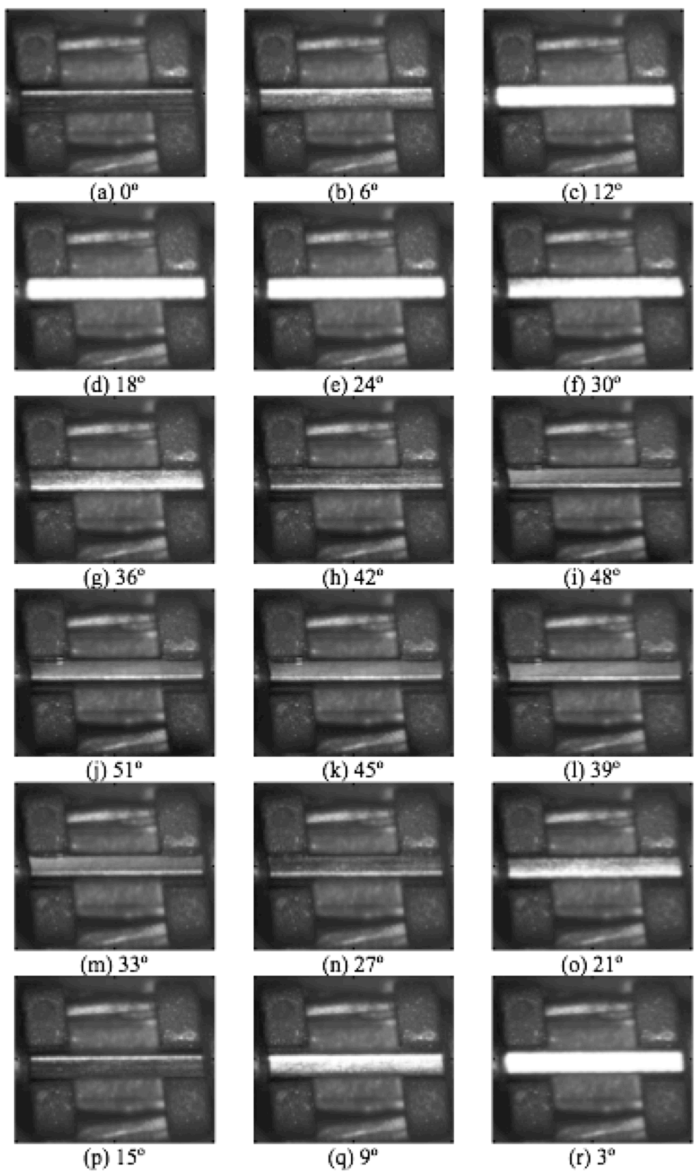

Figure 5.6 Orthos stainless steel bracket displaying bracket slot width deformation

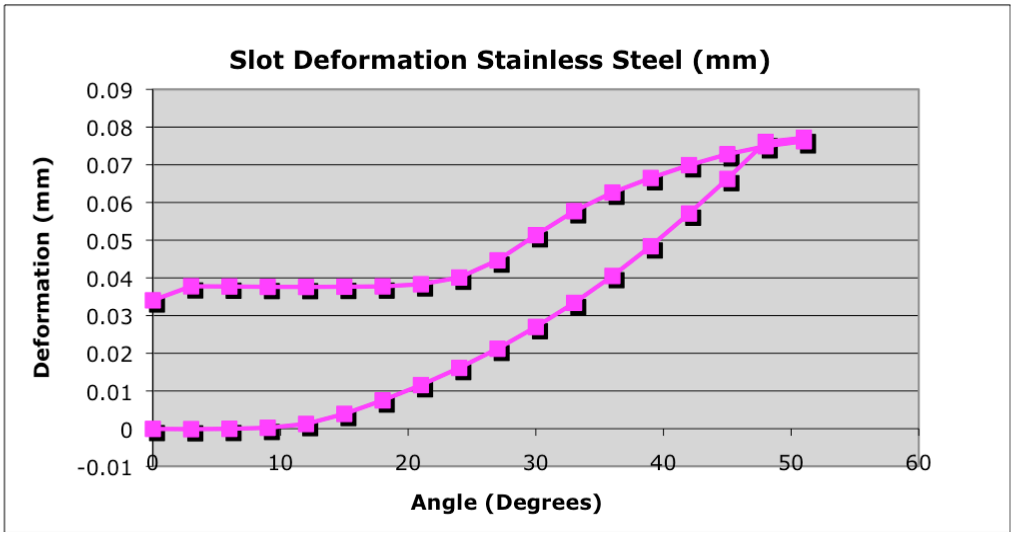

Figure 5.7 Orthos stainless steel bracket slot width deformation (mm)

After the 48° point, the deformation curve begins to flatten, and the torque curve follows a similar change. The leveling off of the slot width deformation curve may signify an increase in wire deformation, reducing the amount of bracket change. The energy produced by the increased archwire torsion would be absorbed in the archwire and its strain, rather than resulting in greater levels of deformation of the bracket

A minor deviation from a linear trend can indicate the beginning of archwire plastic deformation. The torque curve in figure 5.7 appears to be quite continuous in the loading phase, thus making it impossible to estimate a specific stress at which deformation begins.

Throughout the unloading phase neither the bracket torque or slot width deformation is linear in expression. Torque expression decreases dramatically, reaching similar levels as found in the loading phase.

These levels of torque although similar in size differ because they are now expressed at wire angulations that are up to 12° greater in wire angulation as compared to those angles which produced similar levels in the loading phase. For example, if the bracket produced 20 Nmm at a wire angulation of 12° during the loading phase. In the course of the unloading phase, 20 Nmm may be produced at 24°, not at 12° as in the loading phase. This change in torque expression as it relates to wire angulation may result from the interaction of the different materials, hysteresis, varying degrees of plastic and elastic deformation in the bracket and wire, beveling of the wire edge and friction from the wire against the slot walls.

The torque curve appears to flatten at 21° and approximate a horizontal orientation to 0°, depicting the new torque play region. This change in this region is the result of the subsequent plastic deformation that has occurred within the bracket slot and the archwire. Any change within the bracket slot and the archwire, results in a potential change in the prescription that could be realized from reuse of the wire and/or the bracket.

The deformation curve expresses a horizontal trend starting also at 21° in the unloading phase. This signifies the point at which the bracket slot has nearly reached the point of simply permanent deformation. Consequently, the difference between the slot measurements at 0° at the start and the end of the test cycle is the plastic deformation of the bracket in regards to a maximum torsional angulation of 51°. For the Orthos stainless steel brackets, the mean plastic deformation was found to be 0.038 mm, with a standard deviation of 0.016 mm.

Orthos Titanium

The OrthosTi bracket displays a similar shaped torque expression curve (Figure 5.8) as that of the stainless steel version. A mean maximal torque expression of 98.994 Nmm expressed by the OrthosTi was 6.937 Nmm greater than that of the steel bracket and considered to be significant with a p value of 0.002. The OrthosTi brackets though encompassed up to two times greater standard deviations at this angulation and all others that were deemed as significantly different between the two bracket metallurgies.

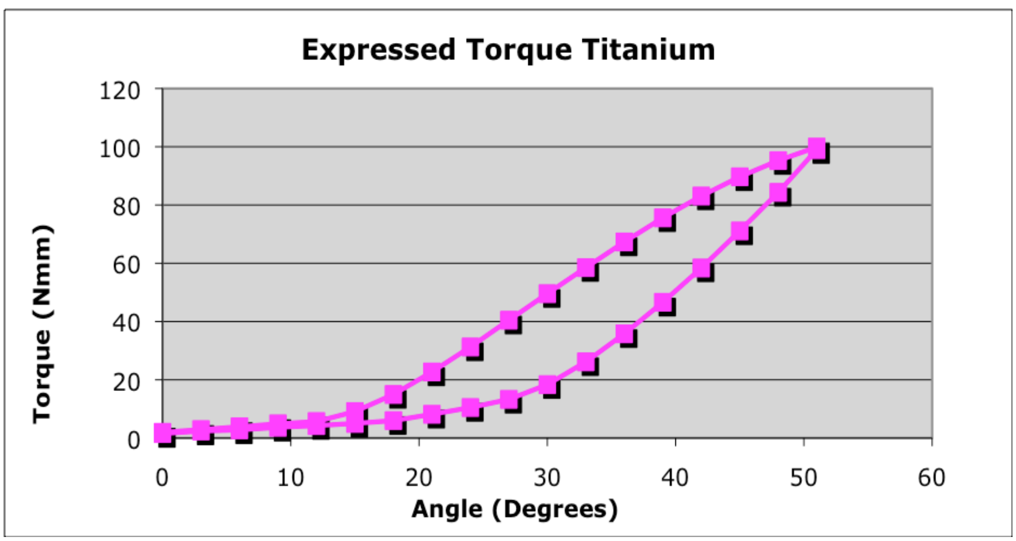

Figure 5.8 Plot of Orthos titanium torque expression (Nmm)

Similar to the steel bracket, the titanium version also expresses a flat torque curve in the loading stage up to the 12° point, whereby it begins a linear ascent up to 45°, at which it starts to flatten like the steel version.

The reason for such a similar curve is likely due to the same reasons as brought forth for the stainless brackets. Which may include potential archwire deformation, bracket deformation and/or the archwire's cross sectional diagonal passing through the perpendicular orientation.

The titanium unloading curve also depicts a linear decrease in torque from 48°rtn to 33° and does not flatten until the 15° point. In addition, it is through the unloading phase that the titanium expresses significantly more torque than the steel version. This difference between steel and titanium throughout the unloading phase only ranges from 4.964 to 6.937 Nmm, which may not be considered clinically relevant even though statistically it is considered a significant difference. It appears that the greatest benefit of the titanium bracket may be more related to its potential to continue torque expression throughout a greater wire angulation range of unloading than a steel counterpart, rather than the actual amount of torque expressed.

The bracket slot width deformation curve of the OrthosTi bracket in Figure 5.9 is comparable in shape to that of the steel version only in the loading phase. Statistically it was found that there was no significant difference between the two bracket types regarding overall deformation. Both brackets exhibit a comparable loading phase shape, although the titanium begins its a linear ascent at a wire angulation of 9°, which is 3° sooner than for steel. The titanium bracket exhibits less maximal slot change with a mean of 0.068 mm, the steel exhibited a mean 0.077 mm.

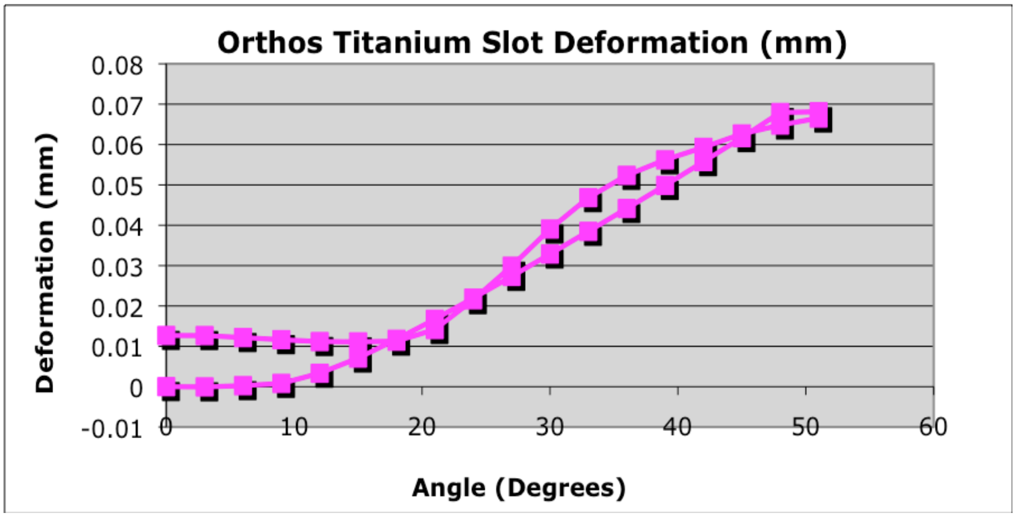

Figure 5.9 Orthos titanium bracket slot width deformation (mm)

Examination of the variance in the sample pertaining to torque expression yields interesting results. At test collection points of 12° and 24° the standard deviations of both the steel and titanium brackets are similar. From collection points at 36° to 36°rtn inclusive, the titanium brackets exhibited almost twice the amount of standard deviation exhibited by steel brackets. It appears then that at higher wire angulations, titanium brackets express greater variation in torque expression than steel. Clinicians using these brackets should be aware of this variance, so as to maintain maximal clinical efficiency when attempting to torque anterior dentition.

Both the steel and titanium brackets presented similar variance pertaining to overall bracket slot deformation. However steel brackets demonstrated more than twice the standard deviation in plastic deformation compared to titanium brackets. The use of the steel bracket may allow greater consistency during torque application. Clinical uncertainty may present after the first torquing cycle since plastic deformation alters the “bracket prescription”.

The difference in this maximal deformation may be attributed to a number of potential reasons. The OrthosTi bracket which uses a Ti-6-4 composition has a smaller modulus of elasticity (114GPa) than 17-4 PH steel (196 GPa).

Young’s modulus is a measure of the stiffness of an isotropic elastic material. As the titanium has a lesser modulus than the steel bracket, it has a lesser degree of stiffness. Titanium also has a greater yield strain than stainless steel which, with the lower modulus, results in a greater ability to return to original shape for similar deformation levels. The steel version having a modulus nearly identical to that of the wire, results in greater amounts of permanent deformation due to its high stiffness and lesser degree of ability to return to its original shape.

It is also possible that the explanation could be related to a geometric shape difference between the steel and titanium versions if it exists. This is specifically regarding bulk mass of material in the area of likely flexure near the base of the slot at both the superior and inferior aspects of the bracket, as greater mass in this area will lead to greater resistance to flexure and potential ensuing deformation.

Shape difference between the stainless steel and titanium, if pertaining to slot width dimension could explain why the titanium brackets expressed greater amounts of torque throughout the complete test cycle. If the titanium brackets encompassed a smaller bracket slot width, they would have a reduced clearance angle, and potentially express greater amounts of torque at similar wire torsion angles. Further to this, if the titanium brackets comprise of larger geometric dimensions this too could potentially explain why they exhibited greater overall torque expression

The unloading phase of the titanium bracket depicts the resilience of the bracket regarding deformation and its inherent properties that minimize the amount of plastic deformation. The unloading curve closely approximates that of the loading phase displaying a period from 45° to 24° whereby the slot dimension is slightly greater in the unloading phase.

The titanium brackets mean plastic deformation was 0.013 mm, 2.9 times less than that of the steel bracket and is considered statistically significant different than the steel version. Therefore, the titanium bracket is likely to hold its geometric shape and consequently its inherent prescription more consistently in treatment because for similar deformation levels, steel will undergo yield before titanium as its yield point is lower. This may provide partial explanation of the titanium bracket's ability to express higher levels of torque during the unloading phase. Therefore, leading to greater overall treatment efficiency.

The steel bracket expressing almost three times the amount of permanent bracket slot width deformation that a titanium bracket exhibits upon a single wire torsion cycle is important to note. Concern then arises if multiple applications of torque were delivered through wire rotation to the same bracket, would steel brackets continue to deform plastically more than a comparable titanium version? Moreover, how does the bracket prescription change with repeated uses in either bracket type? These are important questions for future study.

This study confirms that bracket slot width deformation does occur in the Orthos bracket system. Deformation of the bracket slot width, alters an orthodontic bracket's initial prescription. Modification of the slot width in such a manner leads to decreased clinician control and inefficiency regarding torque expression and/or generation. Bracket slot deformation as it relates widening of the slot on the facial aspect could lead to diminished axial control in the mesial / distal plane, but the greatest affect of this type of slot deformation is regarding diminished torque control.

Although the majority of previous torque studies have had primary focus on torque expression during the loading phase, this study has presented that greater emphasis should be placed upon torque expression during the unloading phase. Expressed torque from the unloading phase mimics the movement of a tooth in the mouth in response to a moment of torque. As a tooth rotates from applied torque, the archwire derotates within the bracket slot, which is the unloading phase. It is this prolonged application of torque after the initial application that has the greatest clinical relevance as it imitates the effect of the torque expression between adjustment appointments. Understanding the expression of torque from different bracket/wire combinations throughout treatment as it occurs between appointment is crucial to clinical proficiency and and overall treatment efficiency.

5.5 Conclusions

The following conclusions were determined regarding the torque expression and concurrent deformation characteristics regarding both the stainless steel and titanium version of the Orthos bracket:

1. Wire angulation and metal type are significant factors regarding both torque expression and bracket slot deformation.
2. The titanium brackets exhibit greater variance regarding torque expression.
3. The titanium brackets express significantly more torque than steel brackets throughout the unloading phase. This difference being less than 7 Nmm, may not be clinically relevant though.
4. Overall deformation encompassing both elastic and plastic bracket slot change is not significantly different between both bracket types.
5. Stainless steel brackets displayed significantly more plastic deformation than the titanium brackets. Steel brackets exhibiting almost 3X more permanent change.
6. Small increases in wire angulation result in large changes in torque expression after the archwire has engaged the bracket slot walls.
7. The use of titanium brackets may result in greater clinical efficiency due to continued torque expression as a tooth moves and greater reliability regarding maintained bracket slot geometry and subsequent torque prescription.

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Chapter 6 General Discussion and Conclusions

6.1 General Discussion

Upon evaluation of current and past research literature, it is clear that investigation of orthodontic bracket torque expression and bracket deformation is lacking. Likely, due to the lack of adequate technology in years past. Moreover, without equipment capable of analyzing these parameters thoroughly and accurately, it is plausible that many of the previous conclusions may be weak in terms of validity.

Through modification of the orthodontic torque measurement device and the addition of powerful custom software this project has developed a means of measuring torque expression and potential concurrent bracket deformation that has yet to be realized in the research and development field of orthodontics. Significantly reducing the potential of human error through automation of the testing procedure and incorporating the previously validated technique of digital image correlation to measure in plane distortion, it is likely that the device is the most technologically advanced of its kind regarding orthodontic bracket torque and deformation analysis.

Examination of the results presented regarding the torque expression of the Orthos steel and Orthos titanium brackets concluded that the titanium brackets do produce greater amounts of torque expression than their steel counterpart. The difference was only significant at the maximal wire torsional angulations and upon the wire's return to a neutral position. Whereas the previous study by Kapur et al. (1999)

found that the titanium brackets produced more torque than a steel bracket only up to 30° of wire rotation.[1] This study has demonstrated that when comparing these two metallurgies, the titanium bracket generates greater torque, with the difference actually increasing at the higher torsional angulations.

Comparing the standard deviations, this study produced opposite findings from that of Kapur et al. (1999).[1] Although the standard deviations between the titanium and steel torque expression followed similar trends throughout the test cycle, titanium brackets presented as much as twice the standard deviation than did the steel version at identical wire angulations.

Exploration of how these two brackets metallurgies responded structurally, particularly the bracket slot width dimension, in response to creating a moment of force brought about very interesting results. It was found that upon initial engagement of the bracket at wire angulations of 12° and 24°, titanium brackets exhibited significantly more deformation. After being tested to a maximum of 51° wire rotation, steel brackets at 12°rtn and 24°rtn test points upon returning to the neutral position expressed significantly more bracket slot width change. Furthermore, the amount of titanium variance was up to 2.6 times that shown by the steel brackets. Difficulty arises in comparing these results to previous studies, because of the lack thereof. This study appears to be the only one of its kind to have been able to evaluate the bracket structure throughout the test cycle. Previous work was limited to initial and final points in the test, being unable to critically assess deformation during torque expression.

6.2 Strengths of the Study

The success of this project is the creation and application of novel in-situ device that can not only measure, but visually depict bracket deformation as it relates to the concomitant expression of a torquing moment. The reduction of the error potential due to human error through the use of automation and computer controlled experimental procedures is a significant advance in the use of the orthodontic torque measurement apparatus. Furthermore, software applications that have been developed, have been created in a manner so as to simplify their use and improve efficiency.

In comparison to previous work with the orthodontic measurement prior to the modifications performed within the scope of this project, data collected in this experiment expresses good uniformity with much smaller standard deviations present in the accumulated data from testing. Moreover, these subsequent modifications have now created endless possibilities for other studies and analysis.

As is often stated, a picture is worth a thousand words, and the images that have been achieved in this pilot work with our novel system our astounding. In conjunction with the appropriate software, depiction of a bracket's structural response to the torquing moment is obvious to the naked eye. Moreover, analysis can now be quickly performed, and potentially requires smaller sample sizes due to the increased accuracy and reliability of the system. Maximizing the automation of the testing procedure has enabled a point whereby any error is very close to being limited to that that is inherent within the hardware itself.

The significant amount of testing with the system in order to improve it to a level of quality that has now been achieved and enabled has enabled planning for future alterations to further increase the accuracy of measurement and its efficiency.

6.3 Limitations of the Study

Great efforts were expended to ensure that this study was performed with attention to thoroughness and accuracy. As with any project, there were limitations that arose:

The most substantial limitation of the study arose from the difficulty in achieving a neutral position for the bracket when setting up for each bracket test cycle. Modifications that included the addition of grub screws to lock the turntables into place and eliminate the slop between components did achieve their intention. This modification led to difficulty in locking the system without creating either a moment or force in any of the three planes. In addition to this, the system does not allow for any rotational adjustment in either the y or z planes of space. If the bracket slot base was not parallel to the sensor, minor forces and moments were created in these planes. This occurred as the elastomeric tie engaged the wire, which is parallel to the load sensor, and forced the wire down into the slot resulting in a wire that would not sit flush in the slot but rather engage part of it due to the nonparallel components.

Another limitation, is that there is no means to view the bracket / wire combination from the side aspect to ensure that at the neutral position, the archwire is parallel and perfectly centered within the bracket slot.

Although the digital image helps center the wire, the lack of focal depth and resultant inability to view the facial and lingual aspects of the slot simultaneously make it difficult to position the wire in a absolute parallel state with the slot walls.

The use of epoxy resin to bond the brackets to the secondary turntable creates potential variability and error as there is no manner in which to determine if and to what amount of flexure is occurring between the bracket and the turntable from epoxy deformation. Other means of attaching the bracket to the turntable were investigated such as welding, but concern arose from potential alteration of the metal properties from such manipulation.

Further limitation arises as any deformation that occurs within the bracket body or pad cannot be accounted for with the use of this device in the manner employed in this study. Imaging is from a frontal aspect only, and to evaluate such potential structural change would entail other means of imaging these areas simultaneously.

Lastly, the elastomeric ligation, presents minor limitation as there is likely variability present within these structures regarding structural integrity, but there is no manner in which to measure each elastomere prior to its use without probable alteration of it properties. This could influence bracket slot deformation, by potentially reinforcing the bracket slot against opening from the rotation of the wire.

6.4 Recommendations for Future Studies

The development of this tool has opened the doorway to a multitude of potential future studies. The possibility to critically evaluate any type of bracket's force and structural response when mated to any type of wire can now be achieved. It would be strongly proposed that future studies examining this would include pretest measurements of the bracket slot dimensions and to include this data in forming any conclusions. By analyzing the bracket slots dimensions prior to testing, statistical examination using such models as the ANCOVA model could be realized and the data scrutinized for a potential Lord's Paradox effect.

Future studies using this device would be best to develop some means of consistently bonding the brackets in a manner that would ensure adequate strength, yet not either alter the bracket's structural properties or create error potential as found with the use of epoxy.

As previously mentioned, the ability of to adjust the bracket position in all three planes would reduce error potential and also significantly increase testing efficiency and efficacy. The incorporation of a fully adjustable (x,y & z planes) stage with the sensor assembly mounted to it, would provide the ability to adjust the bracket position to sit in correct relation to the fixed archwire in the apparatus. The user could then elevate the bracket onto the wire, engage any type of ligation and finish the pretest setup through adjustments in whichever planes are needed to achieve an initial neutral position.

Quite clearly, a number of studies could simply focus on different ranges of maximal torsional angulation, and the testing of the other common archwires such as nickel titanium and beta-titanium (TMA) wires.

Stainless steel wire appears to be the gold standard in studies involving torque and deformation, probably as many clinicians use steel wire in the finishing stages of treatment. TMA is also routinely used in finishing due to its gentler forces and good formability for wire bending. Studies that would investigate the result of using TMA wires, could not only focus on bracket deformation but also perhaps, alteration of the wire geometry after torque application.

Other future studies could entail the investigation of the change in the moment of force generated and subsequent bracket deformation from repeated test cycles on the same bracket and archwire. This would be engaging research, with the capacity of emulating the repeated process often seen clinically. The determination of how the bracket's prescription may change and to what degree from deformation would be extremely relevant research that no one has engaged in to this date. It could prove to be very pertinent in the clinical setting.

In addition to these, studies could include prolonged engagement of an archwire at particular wire torsional angulations which would also emulate a clinical setting. Investigation of both the wire's and bracket's response to a prolonged moment of force, would bring forth information that has yet to be looked upon with such high tech means.

The development of this in situ device provides a remarkable number of legitimate and achievable possibilities for future research.

6.5 Conclusions

There are a number of significant conclusions drawn from this experimental study. This study has brought forth a novel manner in which to efficiently and accurately quantify both torque expression and potential subsequent bracket deformation. It provides the ability to measure the amount and direction of deformation of bracket archwire slot. Unlike previous studies, this device and its accompanying software measures bracket deformation throughout the test cycle and are not restricted to simple comparison of initial and final measurement. Moreover, this study presents for the first time a visual image that clearly displays the change in bracket structure as torque expression is occurring.

Orthondontic brackets particularly the Orthos steel and Orthos titanium brackets do undergo varying amounts of bracket slot distortion and change when a torsional wire angulation of up to 51° is applied within the slot. When comparing brackets of similar design but different metallurgy, the titanium brackets appear to be a better overall choice versus the steel version. Titanium brackets produce greater amounts of torque expression throughout the test cycle range of angulations, yet they express a significant increase over the steel version at clinically relevant angulations. These angulations are at both the initial loading and upon unloading of the bracket. Therefore, it appears as though the titanium bracket retains the ability to continue force delivery even as the tooth would be proclining, leading to greater clinical efficiency.

The downside of the titanium bracket though, is the increased variability in its expression of torque. The steel bracket with overall smaller standard deviations throughout the test range may produce less uncertainty in its use. It could likely instill greater confidence from the clinician regarding regarding the amount of torque produced with a particular angle of engagement.

In regards to torque expression, the determination of which metallurgy is superior, may then depend on the clinician's perspective. If consistency in the amount of expressed torque is precedent, then a steel version is superior. If a greater duration of torque expression is desired, the titanium should be chosen.

When examining the results of the brackets regarding deformation, it is clear that the titatnium bracket is a finer choice as it exhibits less elastic and plastic change. Although initially, the steel bracket undergoes less slot change, the difference is insignificant at this point. It is upon the maximal torsional angulations that the steel exhibited significantly more bracket slot change. For similar great torsional angulations, the steel deformed significantly more.

The steel brackets demonstrated significantly more permanent deformation. As the steel bracket's slot undergoes such change, this introduces greater variability and uncertainty regarding multiple applications of torsion in the wire as is often the case in the clinical setting. Could then the clinician be placing torque in the wire, only to have it lost within the change in the slot from the last activation. Whereas, the titanium bracket demonstrating greater structural resilience, should be more efficient and effective at continued delivery of torque with every reactivation. This statement is reinforced through the observations that the titanium brackets offer less variance in their structural response than do the steel version.

It has been found that wire angulation has the greatest influence on both torque expression and bracket slot deformation.

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