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The effect of wire alloy on torque expression in metallic self-ligating orthodontic brackets

by

Amy Aline Archambault

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Examining Committee

Dr. Paul W. Major, Dentistry

Dr. Carlos Flores-Mir, Dentistry

Dr. Giseon Heo, Dentistry

Dr. Jason Carey, Mechanical Engineering

Dr. David Nobes, Mechanical Engineering

Dedication

I would like to dedicate this thesis to my husband, Paul Morton, who has always encouraged me to reach for my goals. In addition, I would like to thank my parents, Barbara and André Archambault, as well as my sister, Melanie Archambault, for their encouragement and support throughout my education.

Abstract

The objective of this thesis was to identify differences in torque expression between three brands of metallic self-ligating brackets with stainless steel, TMA® and CuNiTi® wire and to compare active and passive designs. 150 of each Damon-3MX (Ormco Inc, Orange, CA, USA), SPEED (Strite Industries, Cambridge, Ontario, Canada) and In-Ovation-R (GAC, Bohemia, NY, USA) brackets were measured for torque expression (Nmm) in an Air-shields Isolette® infant incubator. Torque expression in SPEED and In-Ovation-R brackets began at approximately 7.5° vs. 12° with Damon-3MX. Clinically applicable torque expression (5-20Nmm) was observed from approximately 15-30° in SPEED & In-Ovation-R brackets with stainless steel wire. In Damon-3MX, comparable torsional moments were expressed from 18-30°. With TMA wire, SPEED exerted clinically applicable torsional moments from 18-39°, In-Ovation-R from 21-39° and Damon-3MX from 21-42°. For CuNiTi wire, SPEED expressed torsional moments of 5-20Nmm from 21-72°, In-Ovation-R from 24-63° and Damon-3MX from 27-78°.

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List of Abbreviations

A-NiTi	Austenitic Nickel Titanium
β-Ti	Beta Titanium
CBCT	Cone Beam Computed Tomography
CuNiTi	Copper Nickel Titanium
D	Damon 3MX Bracket
EARR	External Apical Root Resorption
I	In-Ovation-R Bracket
M-NiTi	Martensitic Stabilized Nickel Titanium
NiTi	Nickel Titanium
NS	Not Significant
OMSS	Orthodontic Measurement and Simulation System
PDL	Periodontal Ligament
S	SPEED Bracket
SD	Standard Deviation
SS	Stainless Steel
TAD	Temporary Anchorage Device
TMA	Titanium Molybdenum Alloy or Beta (β)-Titanium

Chapter 1. Introduction

1.1 Terminology

The specialty of orthodontics aims to create ideal smile esthetics and occlusion. This goal is achieved via the movement of teeth through alveolar bone. The control of maxillary incisor root torque permits a normal inter-incisal angle, adequate incisor contact and sagittal adjustment of the dentition in order to achieve ideal intercuspation.[1]

From an engineering perspective, torque is the tendency of a force to rotate an object around its axis.[2] In this document, the term torque moment will be used to describe the moment acting on a tooth to alter buccal-lingual root inclination. Changes in root inclination are achieved through the activation generated by the torsion of an archwire in a bracket slot.[3] Torsion will describe the archwire twist, or activation, that generates a torque moment (Figure 1.1). Torsional load will describe the magnitude of the torsional force acting on the tooth. Torque expression will be employed to describe the ability of a bracket and wire combination to create a torque moment. Synonymous terms for root torque include third order movement and root uprighting.[5] The engagement angle describes the difference between the torsion of the archwire in the bracket slot from a passive position, aligned with the slot walls, to the angle at which two diagonal corners of the archwire make contact with the opposing bracket slot walls.[6] Synonymous terms include play, deviation angle and bracket slot tolerance.

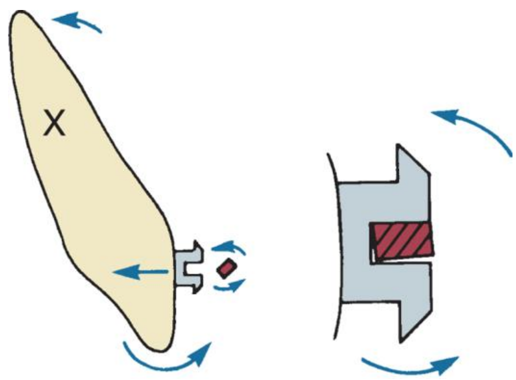

Figure 1.1 Torque Moment in Bracket Slot[4]

1.2 Bracket-Related Factors Affecting Torque Expression

Depending on the magnitude of the torsional load, the dimension and quality of the wire, the "play" available for the wire in the bracket slot, the inclination of the tooth, and the deformity of the bracket, the archwire moves the root of a tooth in a palatal or buccal direction through the tension generated by the torsion in the activated wire. The moment applied to the bracket by a wire in torsion is a research area in orthodontics that has received little attention. This is most likely due to the difficulty in accurately delivering an exact amount of torsional load while measuring the torsion of the wire.[7]

1.2.1 Engagement Angle

The use of a maximum prescription archwire in a pre-adjusted bracket is designed to produce three-dimensional forces to move teeth. These forces result from an intimate fit of the wire in the bracket slot. Any difference between the two will result in incomplete transmission of the bracket prescription to the teeth, periodontal ligament and alveolar bone.[8] Although the engagement angle may be calculated by extrapolating the straight portion of the clockwise and counterclockwise twist-torque curves to the x-axis;[9] a more accurate method involves ascertaining the angle of torsion at which a positive torque moment is first observed.[10]

This concept of engagement angle has been understood for many years; however, it is complex as it involves the interaction of the bracket slot and archwire. If differences due to clinical factors such as initial tooth position, initial tooth movement in the periodontal ligament (PDL) and lip force are eliminated, the discussion can be simplified in order to better understand the interaction of the wire and bracket. From an *in vitro* perspective, alterations in bracket slot design, ligation method (wire or alastic ties in the conventional designs vs. active and passive self-ligating designs), metallurgical properties of the bracket slot, clip/door and wire as well as wire size and shape can dramatically affect torque expression and consequent tooth position.

1.2.2 Bracket Slot Dimensions

Several studies have measured bracket slot dimensions and found them to be oversized relative to the manufacturers' stated values.

Meling et al calculated basic geometries for Ormco's 0.018 inch standard edgewise bracket slot height based on archwire dimensions and cross-sectional shape using a geometrical formula describing the relationship between bracket slot height, wire dimensions, wire edge bevel, and torsional play (engagement angle). One archwire with known dimensions and edge bevel was used for all measurements in order to ensure than any variability was due to the bracket slot size and not the wire dimensions. Results demonstrated that the mean bracket slot height was 0.0187 +/- 0.0001 inches, ranging from 0.0185-0.0189 inches. This range is outside the commonly quoted range for the Ormco standard edgewise bracket.[6]

The slot size of several brands of 0.022 inch slot orthodontic brackets was evaluated in 2004[8]. All bracket slots were found to be oversized; though, three brands (Twin Torque, Clarity and Mini Mono) were within 5% of their stated dimensions with virtually parallel slot walls. The Victory Series slot diverged slightly with the superior aspect of the slot oversized by 6%. The Maxi Advant-Edge, Damon II SL, Elite Mini Opti-MIM Roth, and MBT were all convergent. The base of the Damon II SL slot was oversized by 17%. The Discovery bracket was convergent with the slot base 24% oversized. A 7% difference between the width of the slot at the top and at the base was noted.[8]

There is division in the orthodontic world regarding which slot size is best of the two available dimensions, 0.018 and 0.022 inch. The latter was developed first, for use with gold archwires. When stainless steel archwires were introduced in the 1930's, some clinicians felt that the increased force, in comparison to gold, was detrimental to the oral tissues and the 0.018 inch slot was born. This dichotomy still exists today, most likely due to equal effectiveness of both dimensions.[8]

There is evidence that, despite bracket slot size, slot dimensions are often larger than the manufacturers' claim, which may lead to increased engagement angles in torque applications.

1.3 Clinical Implications

1.3.1 The Biology of Tooth Movement

Orthodontic treatment is founded on the principal that when low magnitude, continuous force is applied to a tooth; blood flow through the PDL is decreased leading to metabolic changes. These changes lead to tooth movement approximately two days later.[10] Heavy sustained pressure occludes the blood vessels of the PDL which results in necrosis, cell death and underlying resorption. Tooth movement occurs 7-14 days later. Light, continuous force is the desired method of tooth movement due to greater efficacy and less pain; however, in reality, even with light forces, small areas of necrosis are likely to develop in the PDL.[4]

Orthodontic force can have a deleterious effect on the dental pulp. Mild pulpitis often occurs after the appliances are activated and lasts for a few days. This type of reaction is not significant in the long term. Conversely, heavy continuous force can lead to loss of tooth vitality.[4]

Roots can also be damaged as a result of orthodontic force.[4] Apical root resorption occurs mainly in the anterior teeth with maxillary lateral and central incisors being affected most severely. Extraction treatment, longer treatment duration and horizontal displacement (retraction) of maxillary central incisors have all been associated with an increased risk of root resorption.[11] Furthermore, a meta-analysis of treatment-related factors identified two factors that were highly correlated with external apical root resorption (EARR); total apical displacement and treatment duration.[12] When applying root torque, an orthodontist moves the root of a tooth or a group of teeth through apical bone in order to achieve treatment goals at the risk of causing EARR. Light forces and modest, rather than significant root movement are therefore important in limiting the risk of EARR.

1.3.2 Torque Expression

Unfortunately, the minimum torque moment required to produce clinically effective torque is not known at this time. Nor is a clinically effective and safe range of torque moments known, with a maximum above which most patients would experience negative effects such as root resorption. Due to the lack of convincing evidence regarding a clinically effective range of torque moments, a range of 5-20Nmm, which has been suggested in the literature[13] will be assumed to be clinically effective.

Further to knowing the range of clinically effective torque moments, it would also be helpful to know the amount of archwire twist that is required to produce a clinically significant range of torque expression for commonly used wire alloys such as stainless steel, titanium molybdenum alloy (TMA®) and copper nickel titanium

(CuNiTi®). Tooth movement in response to archwire twist and the resultant couple will vary depending on patient factors such as age, sex, bone biology and root form; therefore, clinical trials are required to determine this clinically applicable range of torque moments. This information would allow the clinician to predictably alter incisor position with a minimum amount of force, thus minimizing damage to the roots.

Reitan evaluated the histological response of three maxillary first bicuspids to orthodontic force applied with a 50g torquing spring.[14] The relative amount of bony resorption and apposition was noted; however, force levels were measured indirectly. The rate of cuspid tipping in response to the application of "light" and "heavy" forces was measured by Storey and Smith.[15] Springs with known load-deflection characteristics were employed. Tooth movement was measured in relation to a fixed point on the opposing arch when patients had closed in centric occlusion. It was concluded that pressure, not force, was responsible for tooth movement. The maximum rate of tipping was found in the 1.6-2.7 Newton (N) range (160-270 g). The average force needed for optimal cuspid tipping was 180 g. This value corresponds to a force per unit area (pressure) of 197gcm². Burstone et al observed optimal rates of tooth movement in the range of 50-75g (0.5-0.75N) for the retraction of anterior teeth in nine year old children with simple tipping and continuous force application.[16] It was noted that an increase in force did not increase the rate of tooth movement; however, increased pain and mobility were noted.

In 1995, Lee et al[5] evaluated root torque by comparing average weekly changes in axial inclination of teeth to the average torque moment exerted by orthodontic springs in four patients. Tooth inclination was determined by measuring the angle between three points with a protractor: a fixed point on the molar tube (for movement of cuspids and bicuspids) or a fixed point on the central incisor bracket (for movement of lateral incisors), a fixed point at the end of a rod inserted in a vertical tube on the tooth being moved and a fixed point on the tooth to be moved. For the first patient, the maxillary lateral incisor roots were displaced lingually. The maxillary cuspid roots were moved distally for the second and the fourth patient. Finally, the second premolar roots were moved distally for the third patient. Force magnitudes were chosen at random due to a lack of previous studies. The patient whose lateral incisor roots were displaced lingually was treated with different springs for each lateral incisor. On the right side, a 0.016 inch heat-treated stainless steel spring with an initial torque moment of 11.76 Nmm lead to 12.0° of angular change (lingual root movement) after four weeks, without re-activation. On the left side, a 0.014 inch spring with an initial torque moment of 4.85 Nmm produced 5.3° of root uprighting. For a cuspid tooth subjected to a moment of 12.64 Nmm, 9.5° of root uprighting occurred in five weeks; whereas, a 21.95 Nmm moment caused 18.5° of root uprighting in five weeks. Conversely, in another patient, a 31.78 Nmm moment produced 9.0° of uprighting. In the same patient, a 36.7 Nmm moment resulted in 18.0° of uprighting in five weeks. Hence, there is great variability of patient response to such torque moments. The small sample size and lack of control for potentially confounding variables such as tooth type, sex (three male and one female), age, amount of initial force application, rate of spring release and amount and duration of torque exerted limit causal inferences and inferences to the population of all orthodontic patients. A large, controlled and randomized clinical trial is necessary to determine optimal root torque moments.

If 5° is chosen as the minimum required for a clinically detectable amount of change in root inclination and this amount of lateral incisor root uprighting occurred in four weeks with the application of a 4.85Nmm moment, then approximately 5Nmm may be taken as the minimum torque moment necessary to produce clinically significant torque expression. Based on studies of cuspid teeth[5,17] which have larger and longer roots than incisors, a maximum moment of approximately 20-25Nmm may be assumed.

1.4 Studies Measuring Torque Expression *In Vitro*

Several studies have been conducted measuring torque expression in various units using different devices and techniques. In 1977, Steyn et al developed a model for measuring the inclination of the central and lateral incisors as torsion was applied to the archwire. This torque moment was achieved by adjusting the sliding base and steel beam of the apparatus in order to cause the tooth the swivel around the arch. Each tooth had its own pressure-sensing device. A self-balancing indicator pointer which was proportional to the stress and to the torque moment was deflected and the inclination of the tooth was recorded.[18] This study did not account for the engagement angle and the torque moment at each tooth was not calculated;[7] however, it was one of the earliest attempts to quantify torque expression in orthodontic brackets.

Since then, several studies have been conducted measuring torque force and moment, engagement angle, degree of twist at a particular torque moment deemed to be clinically relevant and torsional angle. Dellinger et al[19] and Creekmore et al[20] reported values on the engagement angle between the bracket and wire during wire torsion, based on manufacturers' information. Creekmore used standard wire sizes and a nominal value for bracket slot size derived from a range of manufacturer dimensional tolerances; whereas, Dellinger focused on the manufacturer dimensional tolerances for the wires, using standard slot sizes. Creekmore's engagement angle values were higher than Dellinger's.

In 1982, Hixson et al measured the engagement angle in both new and recycled metal brackets from three companies using the Torquemeter Model #783-C-2 Power Instruments Inc., Skokie, Illinois, USA.[21] They found much greater engagement angles for the new brackets than Dellinger or Creekmore did which he attributed to beveled edges on the rectangular archwires. A study by Sebanc et al in 1984 also demonstrated engagement angles higher than those published by Creekmore and also attributed those differences to beveled-edge wires.[9]

In 1994, McKnight et al studied the effects of simulated torque forces on pre-adjusted orthodontic brackets using an Instron machine. This study used two esthetic brackets and a stainless steel bracket with a 0.022 inch slot and a Roth prescription with a torque value of 12° for the upper left central incisor. This study differed from previous studies because it used a bracket with a torque prescription. The wire was twisted and the torque force was recorded. Values increased with increasing wire torsion and also with increasing wire dimension.[22]

Also in 1994, Odegaard et al evaluated the torsional moments developed in orthodontic applications in vitro using an apparatus constructed from 10mm plates and rods with a plastic crossbar held in place by 2 ball bearings. The engagement angle observed between the wire and bracket ranged from approximately 5° between a 0.018x0.025 inch wire in an 0.018 inch bracket slot to approximately 20º for a 0.016x0.022 inch wire in an 0.018 inch bracket slot as determined by the intercept on the twist-moment curve.[7]

In 1997, Meling et al[23] studied the effects of different sizes and types of square and rectangular stainless steel wire using the same apparatus that was use in the Odegaard et al study.[7] The results demonstrated that variation in cross-sectional dimension and edge bevel leads to variable engagement angle (determined by the intercept on the twist-moment curve). Theoretical play was calculated and differences between the measured and calculated engagement angle were observed due to inaccuracies in wire dimensions. Differences between manufacturers for a given dimension were noted providing evidence of inconsistency in the wire manufacturing process.

In 2000, Fisher-Brandies et al[24] investigated the influence of cross-section, edge geometry and structural hardness on torque transmission between a square wire and bracket. Theoretical and measured engagement angles were calculated. The engagement angle was measured at 0, 1 and 3 Ncm. Wires from five suppliers were tested in various dimensions. The mean engagement angle measured was between 0.8 and 7.5° greater than the theoretical engagement angle. Torque loading increased the engagement angle; whereas, the engagement angle decreased with increasing wire dimension. The cross-section of the wires was measured and revealed that on average, they measured 9.7% in width and 10.7% in height below the nominal value. In addition, the edges were markedly rounded.

In 2002, Gmyrek et al[13] investigated slot deformation and the torque capacity of plastic brackets in comparison with those of a metal bracket in vitro. Two experiments were conducted; an activating experiment measured with a precision lathe and a clinical simulation experiment with the OMSS (Orthodontic Measuring and Stimulation System).[25,26] The torque moment was found to be higher in the activating experiment. Torque was lost in the clinical simulation experiment due to increased engagement angle of the twisted archwire attributed to the adjacent teeth, which were embedded in wax.

In 2004, Harzer et al investigated slot deformation and the equivalent torque capacity of polycarbonate brackets with and without a metal slot in comparison with those of a metal bracket. A clinical simulation experiment was conducted using the same apparatus as the Gmyrek et al study;[13] however a leveled maxillary model was integrated into the OMSS. Results showed that the torque moment was higher with the 0.018x0.022 inch wire than with the 0.016x0.022 inch wire. The moment to force ratio was also higher with the 0.018x0.022 inch wire.[1] These values are similar to the results of the Gmyrek et al[13] study; however, the torque moment of the 0.018x0.022 inch wire was slightly higher.

The recent surge in the popularity and selection of self-ligating orthodontic brackets has resulted in questions regarding torque expression among different self-ligating bracket designs. Two recent studies have evaluated torque expression in self-ligating brackets using stainless steel wire.

In 2008, Morina et al compared two brands of self-ligating metal backets; SPEED (active) and Damon 2 (passive), to conventional metallic, ceramic and plastic 0.022 inch brackets. Lingual root torque of 20° was applied with 0.019x0.025 inch stainless steel archwires in leveled and aligned models in the OMSS.[27] The same device used in the Gmyrek et al and Harzer et al studies in 2002 and 2004.[1,13] Differences in mean torque moment between brackets were not statistically significant; however, the Ultratrimm conventional metallic bracket demonstrated a significantly lower engagement angle than the other brackets at 6.6° vs. Speed at 10.9° and Damon 2 and Discovery at 11.1°.

Also in 2008, Badawi et al evaluated torque expression of four brands of self-ligating metallic brackets using stainless steel wire. A novel device consisting of a multi-axis force-torque transducer with a structure capable of maintaining horizontal and vertical alignment between the wire and bracket was employed. A significant difference in the engagement angle of active and passive self-ligating brackets was noted. The active designs, SPEED and In-Ovation-R, began to express torque at 7.5°; whereas, the passive designs, Smart Clip and Damon 2, began to express torque at 15°. Torque expression in the Speed bracket was relatively constant beyond 35°.[10]

Clinicians have many wire and bracket options available to them for the treatment of malocclusions. To date, no studies have compared torque expression between brands of self-ligating metallic brackets with stainless steel and other wire alloys commonly used for applying root torque, such as β-Ti (TMA) and NiTi or CuNiTi. Niti offers low stiffness and superior range; whereas, β-Ti has good range and excellent formability. β-Ti is a good substitute for stainless steel archwires in the intermediate and finishing stages of orthodontic treatment.[28]

Conventional brackets are still widely used; however, self-ligating bracket designs have recently experienced resurgence in the orthodontic marketplace. Active designs apply pressure to the archwire; whereas, passive designs do not, as long as the teeth are aligned. Examples of active brackets include In-Ovation-R (GAC International, Bohemia, NY, USA) and SPEED (Strite Industries, Cambridge, Ontario). Passive designs include Damon 3MX (Ormco, Glendora, CA, USA) and SmartClip (3M Unitek, Monrovia, CA, USA).

This thesis will examine the effect of orthodontic archwire alloy on torque expression in three brands of self-ligating orthodontic brackets: In-Ovation-R, SPEED and Damon 3MX. It was decided not to test the SmartClip bracket in this thesis since it was shown to perform similarly to the Damon 2 bracket in a previous study.[10]

1.5 Research Objectives

The research objectives for this thesis are listed below.

1. To identify differences in torque expression between three brands of metallic self-ligating brackets with stainless steel wire.
2. To identify differences in torque expression between three brands of metallic self-ligating brackets with TMA wire.
3. To identify differences in torque expression between three brands of metallic self-ligating brackets with CuNiTi wire.
4. To identify differences in torque expression between active and passive self-ligating metallic brackets.

1.6 Null Hypotheses

The null hypotheses of this thesis are listed below.

1. There is no difference in torque expression between three brands of metallic self-ligating brackets with stainless steel wire.
2. There is no difference in torque expression between three brands of metallic self-ligating brackets with TMA wire.
3. There is no difference in torque expression between three brands of metallic self-ligating brackets with CuNiTi wire.
4. There is no difference in torque expression between active and passive self-ligating metallic brackets.

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Chapter 2. Orthodontic Archwire Review

2.1 Engineering Definitions of Structural Properties and Characteristics of Metals

The orthodontic literature differs from the engineering literature in its description of the structural and material properties of metals. Below are definitions of terms and basic formulae used in engineering to describe the behavior of a metal rod in torsion. More details about structural and material properties can be found in such texts as Ugural[1] and Smith.[2]

Shear strength: shear strength is the ability of a material to resist yield (S_{Sy}) or structural failure (S_S) when subjected to shear force applications.[2] This is a material property, independent of cross sectional size or shape.

Shear stress (τ): Direct shear stress is a stress that is applied parallel or tangential to the face of a material. Direct shear stress is the shear force divided by the area over which the shear force acts.[2]

Shear strain (γ): shear strain is the shear displacement divided by the distance over which the shear stress acts.[2]

Elasticity in shear: elastic deformation is said to occur when a metal is deformed by a force but returns to its original dimensions after the force is removed. The shear modulus (G) is defined as the shear stress divided by the shear strain in the linear elastic region of the shear stress-shear strain diagram for a metal.[2]

Torsional stiffness (K_θ): Torsional stiffness is a structural property that determines how a piece of material with specified geometry responds to torsional loading.

Torque, in the linear elastic region of deformation, is related to the angle of twist by:

$$T = K_\theta \theta$$

where T is torque (Nm), K_θ is torsional stiffness and θ is the angle of twist. Torsional stiffness is defined for a beam as:

$$K_\theta = \frac{GJ}{L}$$

Where G is the shear modulus of the material, J is the polar moment of inertia and L is the length of the beam. The polar moment of inertia is a measure of a beam's cross sectional area to resist torsional deformation and is a function of the shape of the cross section.[1]

Torsional shear stress: Shear stress caused by the application of torsional loading.[1]

2.2 Introduction to Properties of Orthodontic Archwires

A basic understanding of the stress-strain characteristics of elastic materials is required in order to select an archwire that will meet specific treatment objectives. An archwire may be considered a beam non-rigidly supported on both ends. When a force is applied, the response can be measured as deflection, in the case of torque, twist.[3] Formulae based on torsion theory suggest that the behavior of an archwire in torsion depends on the following characteristics: cross-sectional shape and size, wire length (interbracket distance), shear modulus and wire material shear yield strength.[4]

Hooke's law applies to the elastic behavior of materials in the linear portion of the load-deflection curve. Austenitic active NiTi (A-NiTi) alloys do not obey Hooke's law due to their non-linear response to loads, allowing a constant force to be applied despite greater load application.[3]

Archwire torsion is assumed to produce a perfect force couple in an orthodontic bracket whose moment will displace the root, producing root torque. Root torque is a combination of bodily movement and rotation. A two-point contact is necessary in order to generate a torque moment.[5] Archwires are designed to move teeth with light, continuous forces.[3] Light forces reduce the potential for discomfort, tissue hyalinization and undermining resorption. Ideally, the archwire should behave elastically for weeks to months after force application. Properties to consider when choosing an archwire include; esthetics, biohostability, friction, formability, weldability, resilience and springback.[6]

2.3 The Effect of Archwire Alloy on Torque Expression

2.3.1 Stainless Steel

Stainless steel archwires were developed after World War I, replacing the gold alloys which had been used for orthodontics up until that time. Stainless steel alloys were strong and their stiffness was 93-100% that of conventional carbon steels.[6] Compared to gold, these alloys had improved strength and rigidity and were equally corrosion resistant.[3] Type 300 stainless steel alloys were used for most orthodontic materials by the 1950's, typically containing 17-25% chromium and 8-25% nickel, with the balance composed of iron.[6] A typical blend of 18% chromium and 8% nickel is used in orthodontics today in a formulation is known as 18-8 stainless steel.[3] Stainless steel has less elasticity than TMA and NiTi. When activated in bending or torsion, stainless steel is capable of storing less energy. This implies that stainless steel wires produce higher forces that dissipate over shorter periods of time, requiring more frequent activations and producing higher forces.[7]

2.3.2 Beta (β)-Titanium

Beta-titanium (β-Ti) archwires were introduced in 1980. Stabilized beta-phase titanium alloys contain approximately 80% titanium, 11.5% molybdenum, 6% zirconium, and 4.5% tin. Pure titanium is approximately half as stiff as stainless steel.[6] This alloy was developed by Dr. Burstone who wanted to produce an alloy with deactivation characteristics less than half that of stainless steel and twice that of martensitic stabilized nitinol.[8] Subsequently, Ormco developed a low-stiffness beta-phase titanium-molybdenum-alloy called TMA®. In comparison with Nitinol™, TMA was smoother and had good weldability and formability. In comparison with stainless steel, TMA produced gentle, linear forces with greater elastic range. These characteristics made it an ideal archwire in many ways, though its high coefficient of friction and consequent inability to allow sliding of teeth were a drawback for space closure.[6] β-Ti has a flexural modulus that is less than half that of stainless steel and almost twice that of Nitinol making it an ideal archwire when force magnitudes between those of stainless steel and Nitinol are required.[7] Its springback is superior to that of stainless steel allowing it to be deflected twice as far as stainless steel without permanent deformation. These wires deliver approximately half the force of stainless steel wires.[9] TMA offers a desirable combination of strength and springiness as well as good formability, thus making it a good choice for auxiliary springs, as well as intermediate and finishing archwires.[3]

2.3.3 Nickel Titanium

2.3.3.1 History

Nickel titanium archwires were introduced in 1972 under the brand name Nitinol ™ (Nickel Titanium Naval Ordinance Laboratory). Nitinol was a stabilized martensitic alloy with a low force per unit of deactivation (low stiffness)[6] that delivered only one quarter of force per unit area of deactivation of stainless steel.[10]

Nickel titanium (NiTi) alloys have two unique properties; shape memory and superelasticity. Like stainless steel, NiTi can exist in more than one crystal lattice structure, austenite and martensite. In the late 1980's, Chinese and Japanese manufacturers developed their own versions of NiTi archwires which were able to take advantage of phase transformation, unlike the original Nitinol alloy. By the mid nineteen nineties, Copper NiTi was introduced in the USA.[3]

2.3.3.2 Superelasticity

When stainless steel or β-Ti archwires are activated beyond their elastic limits, the resulting deformation remains after the mechanical stress has been removed. Below the elastic limit, the strain is proportional to the applied stress, the elastic modulus is constant and the stress-strain curve is linear.

In contrast, superelasticity is the ability of a wire to undergo high levels of elastic deformation. NiTi archwire alloys undergo superelasticity, exhibiting a non-constant elastic modulus which varies according to the amount of activation.[11]

Two nitinol-type alloys are available today that are active, undergoing the shape memory effect: an active austenitic alloy and an active martensitic alloy.[6] Superelasticity can result from a mechanical deformation (in an austenitic active alloy) or from a temperature change (in a martensitic active alloy) which initiates a structural change in the molecules that make up the wire.[11] Unfortunately, Drake et al demonstrated a lack of superelastic behavior in torque applications below 45° of torsion in several brands of wire sold as superelastic, including Ormco CuNiTi 35°C.[12]

2.3.3.3 Pseudoelasticity

When an active austenitic wire is engaged into the slot of a bracket on a rotated tooth, the wire transforms to the lower stiffness martensitic phase. This phenomenon is referred to as stress-induced martensitic phase tranformation. Upon de-activation, the reverse occurs. The second (deactivation) plateau occurs at a lower level of force than the initial (activating) plateau.[6] Kusy demonstrated pseudoelasticity in a diagram (Figure 2.1),[6] where (A-B) and (G-H) represent high modulus austenitic regions. Phase transition occurs in the plateau regions from austenite to martensite (B-C) and from martensite to austenite (F-G). After the phase transformation to martensite, the wire deforms elastically (section C-D). Point E indicates failure of the martensitic-phase wire. If the stress is released before point D (represented by point C'), elastic unloading occurs (section C'-F). At point G, the austenitic structure is completely restored. The term pseudoelasticity refers to the fact that a small portion of the total strain may not be recoverable and is represented by the small difference between points A and H. The slopes of the austenitic moduli on loading (A-B) and unloading (G-H) are three times greater than the martensitic moduli on loading (C-C') and unloading (C'-F). This phenomenon leads to lighter forces being felt by the teeth upon unloading. This allows tooth movement to occur at a low and constant force as long as the wire is activated within the plateau region.[6] The difference in force between the activation (B-C) and deactivation (F-G) plateaus represents an energy loss that is termed hysteresis.[3] This transformation is a mechanical analogue of the thermally-induced shape memory effect. The internal structure of the wire changes in response to stress, rather than an increase in temperature. This is possible because the phase transformation temperature of these alloys is close to room temperature.[3]

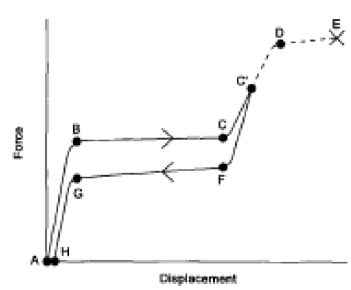

Figure 2.1 Displacement vs. Force Plot Demonstrating Pseudoelasticity[3]

2.3.3.4 Thermoelasticity

Thermoelastic nitiniol-type archwires are martensitic active alloys that exhibit a thermally induced shape memory effect. At high temperatures, NiTi exists in a high stiffness crystalline structure known as austenite; whereas, at low temperatures, the material exists in a low stiffness state known as martensite. A change in structural state occurs at a temperature determined by the chemical composition of the archwire and is reversible.[11] These properties allow an archwire to be engaged into the bracket slots of teeth that are not aligned. As the wire warms to mouth temperature, the wire becomes increasingly stiff which produces forces to align the teeth.

2.3.3.5 Copper Nickel-Titanium

The addition of copper to NiTi alloys lowers rigidity and thus reduces activation and deactivation moments.[13] Ormco produces a 27° Superelastic Copper Ni-Ti wire containing 5-6% copper and 0.2-0.5% chromium. This wire is marketed as an active austenitic wire. The 35° and 40° Superelastic Copper Ni-Ti wires developed by Ormco have higher phase transformation temperatures and thus are considered to be thermoelastic.[6]

Due to the importance of temperature on the elastic properties of thermoelastic NiTi archwire alloys, it is necessary to test under experimental conditions reproducing the intra-oral environment.[14] Filleul observed that upon cooling, Copper Ni-Ti 35° transformed from austenite to martensite at 17.5 °C. At 36.3 °C, the wire transformed from martensite to austenite. At 39 °C, a load of 1190gmm was the minimum mechanical stress required to activate the shape memory effect.[11] Copper Ni-Ti 35° wire can be placed in the mouth without the risk of bond failure due to its low stiffness at room temperature. In the mouth, the unloading force is of a smaller magnitude than the activating force due to its pseudoelastic properties, and remains constant down to 0° of archwire torsion.[11]

2.3.3.6 Formability

Drake et al measured the resilience and elastic properties of stainless steel, β-Ti and NiTi archwires. They noted that NiTi had the largest amount of stored energy and was capable of applying low and constant force; however, it was not easily formable like stainless steel and β-Ti wire.[12] This is problematic when considering torque expression since even with pre-torqued brackets, third order bends (torsion) are sometimes required in the finishing stages of orthodontic treatment. Wire bending (and torsion) is almost impossible with active austenitic Niti because plastic deformation only occurs with extremely high force. These wires must be shaped via heat treatment. In torsion, M-Niti (martensitic stabilized NiTi) must bend more than twice as far as TMA (β-Ti) to deliver the same load, due to its springiness. This is a disadvantage when small and precise adjustments are required. β-Ti or steel wires are a better choice for making final adjustments in tooth inclination.[3]

2.4 A Comparison of the Properties of Orthodontic Alloys

Archwire stiffness modulates the transfer of loads resulting from the activation of a wire engaged in a pre-adjusted bracket slot.[15] If stiffness is known, the torque transmitted to a segment of incisor teeth with a specified amount of third order activation (twist angle) can be estimated. Inter-bracket distance will have a significant effect on torsional stiffness. If bracket deformation is not considered, torque (gm-mm) = (torsional) stiffness (gm-mm/degree) x archwire twist angle (degrees). According to Morina, reduced modulus alloys such as NiTi and β-Ti (TMA) with only a fraction of the stiffness of stainless steel wire may be ineffective in transmitting a torque moment within a bracket slot.[16] This statement differs from the results of a case series conducted by Lee which showed that a 5.3° change in lateral incisor root inclination could be obtained over a period of four weeks with only a 4.85Nmm torsional moment.[17] Though, this was a small, uncontrolled study, Lee demonstrated that higher levels of force are not necessarily better for altering root inclination. Further studies are needed to investigate optimal force/torque levels for altering root inclination.

Kusy et al used formulae to determine the following elastic property ratios of archwires in torsion.[18] A 0.019x0.025 inch TMA wire is 60% as strong, 30% as stiff and has twice the range of a 0.019x0.026 inch stainless steel archwire (0.019x0.025 inch is commonly used today). A 0.019x0.025 inch stabilized martensitic (M-NiTi) wire is 80% as strong, only 10% as stiff and has 5.4 times the range of a 0.019x0.026 inch stainless steel archwire.[19] When comparing 0.019x0.025 inch TMA and M-NiTi archwires, M-NiTi is 1.3 times stronger, half as stiff and has 2.7 times more range.[20] These relationships are summarized in Table 2.1 which was adapted from Proffit describing the relative strength, stiffness and range of 0.019x0.026 inch stainless steel wire, 0.019x0.025 inch TMA and 0.019x0.025 inch M-NiTi archwires in torsion.[3] Alloy, cross- sectional size and gauge length significantly influence elastic stiffness and elastic range of an archwire in longitudinal torsion.[3] Elastic property ratios cannot be calculated for austenitic active archwires because they do not obey Hooke's law.

Table 2.1 Elastic Property Ratios (unitless) of 0.019x0.025 Inch Wire In Torsion

Alloy	Strength	Stiffness	Range
Stainless Steel (0.019x0.026 inch)	1.0	1.0	1.0
TMA	0.6	0.3	2.0
M-NiTi	0.8	0.1	5.4

2.5 The Effect of Wire Characteristics on Torque Expression

Strength is a material property independent of cross sectional shape or length; however, stiffness and range are affected by a change in wire geometry. The performance of an archwire depends on a combination of material properties and geometric factors. Shear stress, rather than bending stress, results from torsion of an archwire. Decreasing the size of a wire decreases its flexural rigidity and increases its range in torsion.[3]

2.5.1 Cross-Sectional Shape

Edge-beveling of orthodontic wires marketed as rectangular can lead to greater engagement angles than theoretical values. Hixson et al measured five sizes of Unitek Hi-T II stainless steel archwires commonly used in finishing in the 0.018 and 0.022 inch bracket slots and examined their edges for beveling.[21] They noted that the engagement angles were 4° higher than the theoretical values stated by Creekmore et al[22] and Dellinger et al.[23]

Sebanc et al measured stainless steel, nickel-cobalt alloy and β-Ti wire from three manufacturers; American Orthodontics (Sheboygan, Wis), Ormco Corporation (Glendora, CA) and Unitek Corporation (Monrovia, CA) with a micrometer that had an accuracy of 0.0001-0.0002 inches. Several readings were taken along each section of wire and averaged. The alternative method of microscopic measurements on cut, polished and mounted specimens was not chosen due to the effect of possible burnishing on the measurements. The average edge bevel contributed to between 0.2-12.9° of the measured engagement angle. β-Ti wires showed the most edge beveling, followed by stainless steel and then by nickel-cobalt. Greater edge bevel led to greater engagement angle. This effect was magnified with greater archwire torsion. The amount of effective torque varied with the manufacturers' tolerances for bracket slots, wire dimensions and edge bevel. The effective torque angle is the difference between the appliance torque prescription and the engagement angle. Due to wire rounding, if an orthodontist uses a 0.019x0.025 inch β-Ti archwire, no effective torque will result unless an angle of at least 22° is incorporated into the bracket-wire combination[24] via bracket prescription, archwire torsion or a combination of both.

2.5.2 Wire Length

Changing the length of an orthodontic archwire will affect its behavior under torsional loading. In torsion, springiness and range increase proportionately with length.[3] This is due to the fact that as length increases, torsional stiffness decreases proportionately based on the following formulae:[1]

$$T = K_\theta \theta$$

$$K_\theta = \frac{GJ}{L}$$

2.5.3 Method of Attachment

Changing the method of attachment of a beam (archwire) will also dramatically affect its stiffness and range.[3] Alastic or metal ligature ties (tight vs loose) and active vs passive self-ligating door/clip designs will alter both the stiffness and range of an archwire.

2.6 Conclusions

The following conclusions can be drawn from this review of archwire properties:

- Torsional stiffness determines how a material will behave in torsion. Torsional stiffness depends on the shear modulus of the material, the polar moment of inertia and the length of the beam.[1] The shear modulus of an archwire depends on the alloy. Stainless steel possesses almost twice the torsional stiffness of β-Ti[6] and 4 times that of NiTi.[10]
- Other wire-related factors such as cross sectional dimension, edge shape, length and method of ligation also affect the behavior of an archwire in torsion.

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Chapter 3. Systematic Review of Torque Expression in Stainless Steel Orthodontic Brackets*

3.1 Introduction

Torque can be defined from a mechanical or from a clinical point of view. Mechanically, it refers to the twisting of a structure about its longitudinal axis, resulting in an angle of twist. Torque is a shear-based moment that causes rotation. Clinically in orthodontics, it represents the buccopalatal crown/root inclination of a tooth and it is an orthodontic adaptation to describe rotation around an x-axis. When applied in an orthodontic archwire/bracket interaction, it describes the activation generated by twisting an archwire in a bracket slot[1]. Orthodontists define torque around the dental arch such that the x-axis follows the curve of the arch. Torque, in this sense, would be rotation perpendicular to the long axis of the tooth. This could be generated by a rotation through a moment or couple of forces. The terms "moment", "torsional moment", "couple", "biomechanical torque" and "third-order torque" appear to be used interchangeably in orthodontic literature to indicate the same loading condition although an understanding of the biomechanical implications of them will not necessarily result in pure torque.

Clinically, torque control is often required in the maxillary incisors for an ideal inter-incisal angle, adequate incisor contact and sagittal adjustment of the dentition in order to achieve an ideal occlusion.[2] A large variation between prescriptions exists with respect to incisor torque values. Maxillary central incisor torque ranges from 12° in the Roth prescription to 22° in the Bioprogressive prescription.[3]

Depending on the magnitude of the torsion, stiffness or resilience of the wire cross section, wire size, edge bevel and manufacturer tolerance, bracket slot size and manufacturer tolerance, engagement angle of the wire in the bracket slot, experimental measurement technique, bracket placement as related to tooth

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morphology,[4,5] and the inclination of the tooth, the archwire moves the root of a tooth through the alveolar bone via localized pressure and tension generated by torsion in the archwire.[2] Most orthodontic treatment is carried out with less than full dimension archwires leading to a lack of cohesive contact between the bracket and the wire known as torsional play or engagement angle.[6]

The current literature on torque expression in orthodontic brackets consists of theoretical models and discussions, *in-vitro* studies employing various measurement devices, brackets and wire dimensions and *in-vivo* studies indirectly measuring torque expression via tooth inclination. It is difficult to predict the amount of torque expression that a clinician can expect from a given bracket and archwire combination. This systematic review was intended to evaluate the quantitative effect of varying the slot size of stainless steel orthodontic brackets and the dimension of stainless steel wire on torque expression. It is our intention to help the clinician to better understand the variables involved in generating torque moments when selecting stainless steel archwires for torque expression.

3.2 Materials and Methods

See Appendix A.

3.3 Results

Thirty-seven abstracts met the initial inclusion criteria. Once the full articles were retrieved, only nine met the final inclusion criteria. Reasons for exclusion at the final selection stage are stated in Table 3.1. A hand-search of the reference lists of the nine articles that met the final inclusion criteria identified two additional articles. A total of eleven articles met the final inclusion criteria (Figure 3.1). Two studies[2,7] were intended to evaluate torque expression in plastic brackets but were included because they used metal brackets as a control. The study by Hixson et al [8] was intended to evaluate changes in bracket slot tolerance following recycling of metal brackets.

Since baseline values using new metal brackets were measured, this part of the data was included in this systematic review. Two articles in Chinese[9,10] were excluded due to language as no translator for the dialect was available.

Table 3.1 Articles Not Selected From the Initial Abstract Selection List With Reasons for Exclusion

Article*	Reason Excluded
Andreasen[1]	Non-stainless steel wires, in vivo study
Bachmann[2]	Graphic data lacking detail
Bai[3]	Language: Chinese
Bantleon[4]	Only auxiliary wires used for torquing
Foglio[5]	Did not measure torque
Foglio[6]	Did not measure torque
Broadbent[7]	Did not measure torque expression
Cassarino[8]	Only auxiliary wires used for torquing
Flores[9]	Did not measure torque expression
Engel[10]	Grey literature (thesis abstract)
Gioka[11]	Review article
Gurgel[12]	Non-stainless steel wires
Isaacson[13]	Did not measure torque expression
Kapur-Wadhwa[14]	Review article
Kesling[15]	Only auxiliary wires used for torquing
Knosel[16]	In vivo study
Meling[17]	Non-stainless steel wires
Meling[18]	Concurrent second-order activation
Meling[19]	Non-stainless steel wires
Nasiopoulos[20]	In vivo study
Owen[21]	Did not measure torque expression
Pandi[22]	In vivo study (clinical trial)
Rosarius[23]	In vivo study
Siatkowski[24]	Did not measure torque expression
Steyn[25]	Did not account for "play," does not measure pure torque
Ugur[26]	Did not use a torque-measuring device (measured faciolingual crown inclination on models)
Wagner[27]	Measured torsional stiffness, not torque expression (wire property only)
Zhang[28]	Language: Chinese

*See separate reference list in Appendix B.

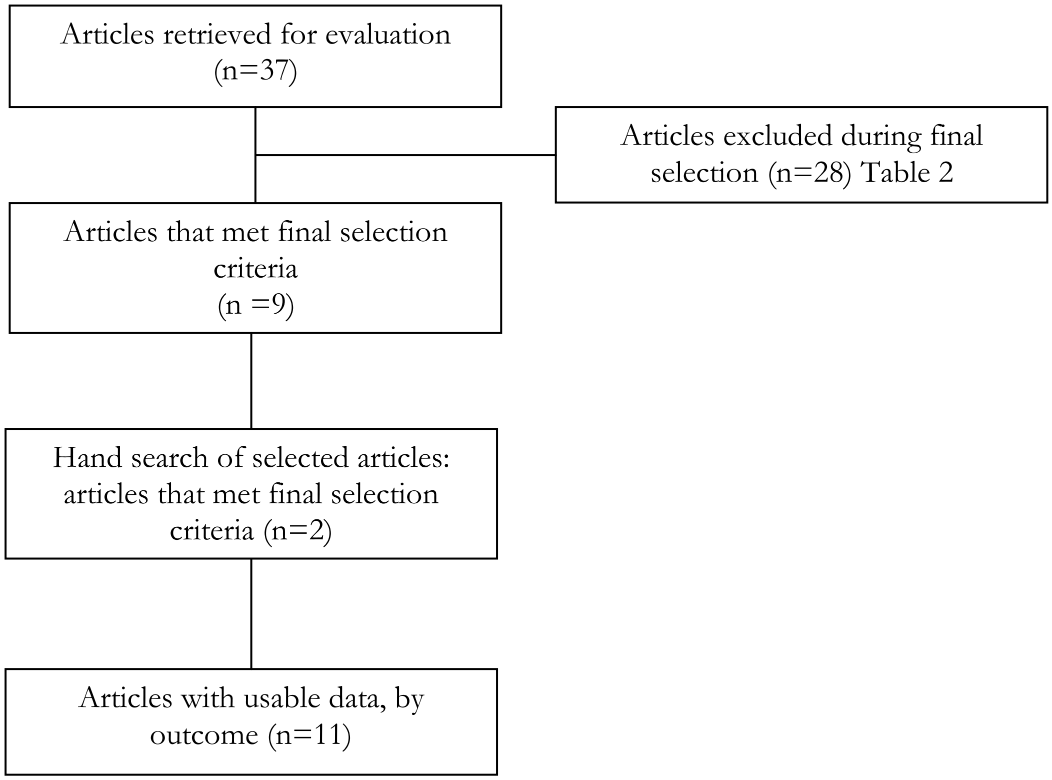

Figure 3.1 Flow Diagram of the Literature Search

The methodology and results of each selected article have been summarized in Table 3.2. There are several variables measured to represent torque expression; however, the results measuring the engagement angle will be emphasized.

Table 3.2 Methodology and Results of Selected Articles

Author	Measurement Device	Variable Measured	Error Measurements	Bracket	16x16	16x22	17x22	17x25	18x22	18x25	18x18	19x25	21x25	21.5x28
Fischer-Brandies[15]	Fixed bracket, design not specified Use of ligatures not mentioned	Torque play (degrees) at 0 Ncm of torque loading[I]	Control measurements were made for torque application.	Ultratrimm (1) Dentaurum edgewise brackets (0.018 slot ss) upper central incisor Inter-bracket distance: not specified	29.3	17.5		12.3						
Gmyrek[7]	Clinical Simulation Experiment: OMSS (Orthodontic Measuring and Simulation System) Wire ligatures were used	Maximum torquing moment (Ncm) at 20 Degrees of labial crown torque[II]	Each bracket-archwire combination was measured five times.	MINI-MONO (2) (0.018 slot ss) upper central incisor Inter-bracket distance: 6mm		1.21			1.94					
Gmyrek[7]	Activating Experiment: Precision lathe. Thermal cycling in saliva substitute. Pre-shaped ligature wires used	Maximum torquing moment (Ncm) up to 30 degrees of archwire twist[II]	See above	MINI-MONO (2) (0.018 slot ss) upper central incisor Inter-bracket distance: 6 mm		2.09			2.80					
Harzer[2]	OMSS Wire ligatures were used	Median of maximum torquing moment (Ncm) at a 20 degree labial crown torque[III]	See above	MINI-MONO (2) (0.018 slot ss) maxillary left central incisor Inter-bracket distance: 6 mm		1.17			2.22					
Harzer[2]	See above	Moment/torque ratio,(Nmm/degree)[III]	See above	See above		0.54			1.24					
Morina[13]	OMSS Wire ligatures were used for non self-ligating (sl) brackets	Mean maximum torquing moment (Nmm) at 20 Degrees of labial crown torque	Each bracket-wire combination was measured 5 times	Speed (3) Damon 2 (4) Ultratrimm (1) Discovery (5) All 0.022 inch slot Inter-bracket distance: 6 mm								8.0 (3.7) 7.8 (4.0) 12.3 (5.5) 7.5 (3.3)		

Table 3.2 Methodology and Results of Selected Articles Continued

Author	Measurement Device	Variable Measured	Error Measurements	Bracket	16x16	16x22	17x22	17x25	18x22	18x25	18x18	19x25	21x25	21.5x28
Morina[13]	See above	Torque loss	See above	See above								10.9 (2.1) 11.1 (2.9) 6.6 (4.7) 11.1 (2.9)		
Odegaard[12]	Novel plate/rod/crossbar apparatus Elastic ligatures were used	Intercept linear stress/strain curve Play = Intercept/2[II]	Total: error measurements taken. Each type of wire was tested with 5 brackets.	Five Ormco (6) wide twin Siamese brackets with (0.018 slot ss) zero torque and angulation Inter-bracket distance: 7 mm		20.4 10.2	14.4 7.2			5.3 2.52				
Odegaard[12]	See above	Degree of twist for a torsional load of 16.25 Nmm (degrees)	See above	See above		26.3 ±0.74	19.8 ±1.05			9.2 ± 0.57				
Meling[11]	Further development on instrument used in Odegaard study[12] Use of ligatures not mentioned	Torsional play based on intercept with deformation axis (degrees)	Correct orientation of brackets was tested. 400 double measurements were taken at 10 Nmm.	Standard edgewise brackets (0.018 slot ss) (tooth not specified) Inter-bracket distance: 4 mm	31.2 (28.8-33.8)	18.5 (16.6-20.4)	11.9 (10.4-14.5)	9.1 (7.5-11.5)			7.2[II]			
Meling[11]	See above	Angle of twist at 20 Nmm (degrees)	See above	See above	45.2 (42.9-50.3)	26.3 (24.6-29.2)	19.2 (17.4-20.7)	14.8 (13.6-18.3)		9.7 (8.3-12.0)	14.6[II]			
Badawi [6]	Novel device with a multi-axis force/torque transducer	Moment of the couple (torque moment) at 24 degrees of torsion (Nmm)	Five Damon 2 (4) brackets and wire measured 10 times each at 4 angles of torsion	Self-ligating maxillary right central incisors: Damon 2 (4) In-Ovation-R (7) Speed (3) Smart clip (8) Inter-bracket distance: 5mm								Damon 2: 5.5 (3.9) In-Ovation-R: 12.4 (6.9) Smart Clip: 6.5 (5.4) Speed: 11.9 (5.2)		

Table 3.2 Methodology and Results of Selected Articles Continued

Author	Measurement Device	Variable Measured	Error Measurements	Bracket	16x16	16x22	17x22	17x25	18x22	18x25	18x18	19x25	21x25	21.5x28
Badawi[6]	See above	Moment of the couple (torque moment) at 36 degrees of torsion (Nmm)	See above	See above								Damon 2: 23.2 (9.6) In-Ovation-R: 27.8 (12.5) Smart Clip: 21.0 (9.5) Speed: 22.4 (6.7)		
Badawi [6]	See above	Torsional play (degrees)	See above	See above								7.5 (Speed, In-Ovation-R) 15.0 (Damon 2, Smart Clip)		
Sebanc[4]	Commercial Torque Meter model 783-C-2 Power Instruments Inc., Skokie, Illinois Use of ligatures not mentioned	Deviation angle (degrees)= torsional play	Several readings were taken along each segment of straight wire.	Maxillary right central incisor brackets (9, 6,10) prescription zero, in 0.018 and 0.022 slot. Inter-bracket (tie-wing) distance: approximately 3mm	21.7 (17.2-24.9)	13.9 (10.3-14.1)		7.05 (6.2-8.1)		18.0 (16.6-20.1)		12.4 (10.9-23.7)	6.13 (4.5-8.3)	
McKnight[5]	Instron machine Use of ligatures not mentioned	Torquing force = force vs angular deflection (g)	None reported	0.022 Roth prescription "A" company (11) ss bracket maxillary left central incisor (torque: 12 degrees) Inter-bracket distance: 8 mm						7.60 (2.80) 16.50 (5.80) 30.90 (9.61) 48.30 (13.71)		8.64 (4.04) 17.30 (7.56) 34.70 (15.21) 57.70 (26.35)	22.30 (4.82) 45.93 (8.20) 87.67 (12.02) 151.83 (15.31)	
Feldner[14]	Torque transducer, brackets bonded to porcelain teeth held by dye stone Elastic ligatures used	Torque in millivolts converted to gm-cm with a calibrated conversion chart at 5 degree intervals beginning at 5 degrees	None reported	Mini Diamond (12) (0.022 inch) edgewise upper right central incisor Inter-bracket distance: 6 mm										41.8 (21.8) 126.4 (25.4) 219.4 (28.9) 315.7 (31.8) 410.8 (33.4) 499.4 (35.5)

Table 3.2 Methodology and Results of Selected Articles Continued

Author	Measurement Device	Variable Measured	Error Measurements	Bracket	16x16	16x22	17x22	17x25	18x22	18x25	18x18	19x25	21x25	21.5x28
Hixson[8]	Torque meter Model #783-C-2 Power Instruments Inc., Skokie, Illinois (adapted) Use of ligatures not mentioned	Torque play (degrees)	None reported	(0.022 slot ss) standard edgewise mandibular canine (11) maxillary lateral (9) maxillary lateral (6) Inter-bracket distance: not specified									8.4 (1.0) 9.9 (1.4) 8.7 (1.5)	

I Value for the Standard rectangular wire (3M/Unitek), Straight arch (Forestadent) and Edgewise rod wire (Ormco), other wires slipped in slot therefore, no torque transmission was possible. Torque play was averaged for five brands of archwires per dimension; however, standard deviations could not be averaged.

II No standard deviation (range) recorded.

III Standard deviations in graphic form in the article.

Key for orthodontic bracket manufacturers:

1. Ultratrimm ® (Dentaurum, Ispringen, Germany)
2. MINI-MONO (Forestadent, Pforzheim, Germany)
3. SPEED (Strite Industries, Cambridge, Ontario, Canada)
4. Damon 2 (Ormco Corporation, Orange, CA, USA)
5. Discovery sl (Dentaurum, Ispringen, Germany)
6. (Ormco Corporation, Glendora, CA, USA)
7. In-Ovation R (GAC, Bohemia, NY, USA)
8. Smart Clip (3M Unitek, Monrovia, CA, USA)
9. (American Orthodontics, Sheboygan, Wis, USA)
10. (3M Unitek, Monrovia, CA, USA)
11. ("A" Company, Johnson & Johnson, CA, USA)
12. Mini Diamond (Ormco Corporation, Glendora, CA, USA)

The angle of twist of the archwire in degrees at 20 Nmm was measured by Meling et al[11] and Odegaard et al[12]. The maximum torquing moment in Ncm at 20° of archwire torsion was measured by Gmyrek et al[7] and Harzer et al[2]. Morina et al[13] measured the maximum torquing moment in Nmm at 20° of archwire torsion as well as torque loss. Results indicate that Morina et al[13] noted mean moments of 8.0 Nmm for the Speed bracket (Strite Industries, Cambridge, Ontario, Canada) and 7.8 Nmm for the Damon 2 bracket (Ormco Corporation, Orange, CA, USA) with a 0.019x0.025" stainless steel wire in a 0.022" slot with the OMSS. Testing the same wire and bracket slot combination at 24° of archwire torsion, Badawi et al[6] recorded 5.5-12.4 Nmm depending on bracket brand, with Speed (Strite Industries, Cambridge, Ontario, Canada) measuring 11.9 Nmm and Damon 2 (Ormco Corporation, Orange, CA, USA) measuring 5.5 Nmm. Gmyrek et al's experiments with the precision lathe produced results that resemble those of Badawi et al[6]. At 30° of torsion, Gmyrek[7] et al reported a moment of 2.09 Ncm (20.9 Nmm). Badawi et al[6] measured the moment of the couple at 24 and 36 degrees with values of 5.5-12.4 Nmm at 24 degrees and 21.0-27.8 at 36 degrees. In addition, Badawi et al[6] measured the engagement angle from the archwire twist to moment graph as the point at which a positive torque moment was detected after the initial lag period. The torque force in g was measured at 5 degree intervals by McKnight et al[5]. The torque moment was measured in g-cm at 5 degree intervals by Feldner.[14] Fischer-Brandies[15] measured the engagement angle in degrees at 0, 1 and 3 Ncm of torque loading. The values chosen to be evaluated in this systematic review were those measured at zero Ncm to represent the contact angle. The engagement angle was measured in degrees. In a 0.018" bracket slot it ranged from 31 degrees with a 0.016x0.016" archwire to 4.6 degrees with an 0.018x0.025" archwire. In a 0.022" bracket slot, the engagement angle ranged from 18 degrees with an 0.018x0.025" archwire to 6 degrees with a 0.021x0.025" archwire.[4,6,8,11,12,15]

3.4 Discussion

The current systematic review identified eleven *in-vitro* studies in which conventional and self-ligating metal brackets that quantified torque expression were tested under different working conditions and different torque parameters were measured. These studies were dissimilar in the methodology and parameters tested. In order to draw a meaningful conclusion, studies measuring the same parameter were compared.

3.4.1 Engagement Angle

The engagement angle was selected for comparison between studies because it was the parameter tested by the majority of the studies. The engagement angle was measured directly in the study by Fischer-Brandies et al.[15] On the other hand, the engagement angle can be measured indirectly from the twist/moment curve using several different methods. Archwire torsion can be measured when a positive torque moment is first observed, as did Badawi et al.[6] This is probably the best method since it uses real data points rather than extrapolated points to identify the angle at which a positive torque moment is produced. The engagement angle can be estimated by extrapolating the linear portion of the twist/moment curve back to the x-axis.[11,12] Alternatively, the engagement angle can be measured by converging the data from both clockwise and counterclockwise torsion and finding the midpoint between the two x-intercepts.[4,8] Extrapolating from the linear portion of the twist/moment curve assumes a relationship that does not exist since there is a "lag" prior to the critical contact angle when the archwire is not engaged in the bracket slot. This method does not account for the twist from the time the wire engages the bracket to the point at which the relationship between the angle of twist and the moment is linear. Rather, it is assumed that the relationship is linear immediately.

Fischer-Brandies et al[15] applied three torque values (0, 1 and 3 Ncm). Results at 0Ncm were utilized in this systematic review because this loading condition represents the critical contact angle. One Ncm was deemed to be clinically relevant whereas 3 Ncm was deemed to be an excessive amount of torque from a clinical point of view; however, it was designed to test the weaknesses of the slot wire system. Since the value at which torque becomes clinically relevant is unknown, this method may be arbitrary. Gmyrek et al[7] has suggested that the range of clinically effective torque is between 5 and 20 Nmm.

3.4.2 Experimental Measurement Devices Used to Quantify Torque Expression

A total of eight devices were used to measure torque expression in its various forms in the eleven studies which met the final inclusion criteria. Several studies measured torque expression with different styles of lathes. [5,7,8,14] Inaccuracy may occur if the pulley does not fit tightly around the lathe, producing an axial force. To prevent this axial force development and consequent energy loss, a pulley that exerts a force couple could be used. In addition, the wire may distort or twist within the lathe leading to frictional torque loss.

Gmyrek et al[7], Harzer et al[2] and Morina et al[13] used the orthodontic measuring and simulation system (OMSS)[16] to measure the maximum torquing moment. This device has 6° of load measuring freedom. It ensures that only torque in one plane is present (buccal-lingual) by automatically adjusting itself through a load sensor feedback system. Torque values were smaller for the OMSS experiments by Gmyrek et al and Harzer et al than for Gmyrek's activating experiment.[2,7] Torque loss was attributed to increased play caused by adjacent bracketed teeth embedded in wax, as opposed to other *in-vitro* studies, such as the activating experiment by Gmyrek et al,[7] in which a device firmly clamps a test bracket or series of brackets on both sides. Badawi et al[6] developed a novel apparatus also using a 6 degree of freedom multi-axis force/torque transducer. Torque was measured as the wire was twisted; all other forces and moments were kept as close to zero as possible with device alignment. Vertical and horizontal alignment was maintained between the wire and bracket during this process. The apparatus consisted of a wire-support substructure and alignment dies on either side of a bracket on a dual turntable system secured over the sensor.

Three articles measuring the engagement angle increased archwire twist incrementally;[6,11,12] however, another two[4,8] only took four data points for each clockwise and counter-clockwise archwire twist and assumed a linear torque-twist relationship extrapolating back to the x-axis to calculate the engagement angle. The collection of more data points would have provided a more accurate curve.

3.4.3 Other Factors Affecting Measured Torque

According to several authors, the mean engagement angle measured was greater than the theoretical engagement angle due to undersized wires with rounded edges.[4,11,15] Another factor that can increase the engagement angle is bracket slot dimension. When torque is applied, notching of the slot walls and additional widening of the slot by up to 0.016 mm can occur.[15] Elastic bracket deformation does not increase the engagement angle but does affect the amount of torque delivered by the archwire twist.

In a study evaluating the effect of elastic ligatures on the engagement angle in Ormco 0.018 inch brackets, elastic ligatures were found to reduce the engagement angle, allowing small torsional moments to develop at lower torsional angles than brackets without elastic ligatures (8.5° vs 18.5°). These small torsional moments (<5Nmm) are not likely to be clinically significant. At 21.5°, the same bracket expressed approximately 4Nmm of torque with or without elastic ligatures. Thus, this effect of limited duration.[12]

3.4.4 Torque Expression in Self-Ligating Metal Brackets

Two recent studies have tested torque expression in metal self-ligating brackets.[6,13] A comparison may be drawn between the results of these studies which both tested

0.019x0.025" wire in the 0.022" bracket slot; however, Badawi et al[6] measured the moment of the couple at 24° of labial crown torque, whereas Morina et al[13] measured the moment of the couple at 20° of labial crown torque. Morina et al[13] noted no significant difference between the moments generated by the Speed and Damon 2 brackets whereas Badawi et al[6] found a significant difference between the two brackets. The two active self-ligating brackets in Badawi's study, In-Ovation-R (GAC, Bohemia, NY, USA) and Speed showed similar results to the OMSS studies by Gmyrek et al[7] and Harzer et al;[2] whereas, those of the passive self-ligating brackets, Damon 2 and Smartclip (3M Unitek, Monrovia, CA, USA) demonstrated lower torsional moments. Lower torque values would be expected in the OMSS experiments due to torque loss due to adjacent brackets; however, in the study by Badawi et al,[6] Damon 2 exhibited a lower moment than in the study by Morina et al.[13] Though, if the standard deviations are considered, this difference is minimized. A difference between active and passive self-ligating bracket designs was evident. If the range of clinically effective torque is considered to be between 5-20 Nmm,[7] this range can be attained with 15-31° of torsion with the active self-ligating brackets and with 22.5-34.5° with the passive self-ligating brackets with a 0.019x0.025" stainless steel archwire in a 0.022" slot.[6]

When the results of torque tests on multiple wires from different companies were averaged, there was no statistically significant difference in the engagement angle between the 0.018" slot and the 0.022" slot for the wire sizes tested (0.018" slot: 0.016x0.016", 0.016x0.022", 0.017x0.025"; 0.022" slot: 0.018 x0.025", 0.019x0.025", 0.021x0.025").[4] Six studies[4,6,8,11,12,15] measured the engagement angle and therefore can be compared, though differences in wires tested and bracket slot dimension tested limit the comparison.

3.4.5 Clinical Relevance of the Results

Clinical research is needed to determine the average amount of tooth movement in degrees produced by the range of torsion resulting in 5-20 Nmm of torque expression. With this information, the clinician will be able to calculate the amount of torque moment required in a particular clinical situation. Standardized bracket slot and archwire dimensions as well as edge bevel are required in order to accurately predict torque expression.

Overall, these studies indicate that the engagement angle is clinically significant and variable and is affected by bracket slot, archwire dimensions and edge bevel. This variable torsional play was found to be larger than the theoretical nominal values published previously by Dellinger et al[17] and Creekmore et al[18]. In a 0.018" bracket slot, the published nominal values are 9.62° according to Dellinger et al[17] and 16.7° according to Creekmore et al[18] for a 0.016x0.016" stainless steel archwire and 1.50° according to Dellinger et al[17] and 2.0° according to Creekmore et al[18] for an 0.018x0.025" archwire in contrast to the values found in this systematic review of 31° and 4.6° respectively. In a 0.022" bracket slot, the published nominal values for play are 11.02° according to Dellinger et al[17] and 14.8° according to Creekmore et al[18] for a 0.018x0.025" stainless steel archwire and 1.74° according to Dellinger et al[17] and 3.9° according to Creekmore et al[18] for a 0.021x0.025" archwire; whereas, this systematic review found the values to be 18° and 6° respectively. In light of these findings, clinicians should consider the magnitude of the engagement angle when selecting bracket prescription and/or torque to be added in the finishing stages.

This systematic review did not attempt to compare self-ligating brackets to conventional brackets. Nine out of the eleven selected articles measured torque expression in conventional brackets; whereas, only two measured torque expression in self-ligating brackets and one study[13] measured both.

From a clinical perspective, initial proclination or retroclination of the anterior teeth and/or buccal or lingual crown inclination of the posterior teeth will affect the engagement angle and the final position of the teeth. This produces variability in the clinical response to a given archwire and bracket combination.

3.5 Conclusions

- The measured engagement angle is larger than theoretical values and is highly variable.
- For conventional stainless steel orthodontic brackets with a 0.018" stainless steel bracket slot, the engagement angle ranges from 31° with a 0.016x0.016" stainless steel archwire to 4.6° with a 0.018x0.025" stainless steel archwire. In a 0.022" stainless steel bracket slot, the engagement angle ranges from 18° with a 0.018x0.025" stainless steel archwire to 6° with a 0.021x0.025" stainless steel archwire.
- Active stainless steel self-ligating brackets demonstrate an engagement angle of approximately 7.5°, whereas the passive stainless steel self-ligating brackets show an engagement angle of approximately 14° with 0.019x0.025" stainless steel wire in a 0.022" stainless steel bracket slot. Clinically effective torque can be achieved with archwire torsion of 15-31° for active self-ligating brackets and with 22.5-34.5° for passive self-ligating brackets with 0.019x0.025" stainless steel wire in a 0.022" stainless steel bracket slot.[6] This difference is due in part to a larger engagement angle in the passive self-ligating brackets compared to active self-ligating brackets.

3.6 References

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Chapter 4. A Comparison of Torque Expression in Metallic Self-Ligating Brackets with Stainless Steel, Titanium Molybdenum Alloy and Copper Nickel Titanium Wires

4.1 Introduction

In order to correct malocclusions associated with irregular axial inclination of teeth, controlled root movement is required. This movement is commonly referred to as third order movement, root torque or root uprighting.[1] An ideal anterior crown inclination contributes to normal overbite and posterior occlusion. When teeth are overly upright, they lose their functional harmony and over-eruption results. When the anterior teeth are upright, the posterior teeth move mesially and spacing results, which can be mistaken for tooth size discrepancy.[2] Correct inclination of the posterior teeth is necessary in order to achieve ideal cusp-to-fossa relationships of the maxillary and mandibular posterior teeth.[3]

The twisting of an edgewise wire in a bracket slot generates a torsional load.[4] Since most orthodontic treatment employs undersized archwires, a lack of rigid control exists between the bracket slot and the wire. This phenomenon is known as the engagement angle, or "play", and may be defined as the amount of rotation, in degrees, that a rectangular or square archwire must be twisted in order to engage the bracket or tube and create biomechanical torque expression. The effective torque angle is the difference between the appliance torque prescription and the engagement angle.[5] Optimal tooth positioning depends upon optimal control of the interaction between the bracket and wire and therefore the closeness of the fit is especially important in torque applications.[6]

Torque expression is the result of the interaction of many factors. Bracket design, wire/slot play (engagement angle), mode of ligation,[7] bracket deformation,[8] wire stiffness,[3,7] magnitude of wire torsion and dimension[8-10] as well as wire edge beveling[5,6,11] have all been suggested as factors that affect torque expression. Clinical factors such as initial tooth inclination,[8] bracket position[12] and tooth anatomy[13,14] are also determinants of torque expression.

Low-modulus alloys such as NiTi result in decreased torque expression relative to stainless steel.[3] Reduced modulus alloys such as NiTi and β-Ti (TMA) with only a fraction of the stiffness of stainless steel wire may be ineffective in transmitting a torque moment within a bracket slot.[7] This statement differs from the results of a case series which showed that a 5.3° change in lateral incisor root inclination could be obtained over a period of four weeks with only a 4.85Nmm torsional moment.[1] Though, this was a small, uncontrolled study, it was demonstrated that higher levels of force are not necessarily better for altering root inclination. Further studies are needed to investigate optimal force/torque levels for altering root inclination, though, it has been suggested that applied torque moments should range from 5-20Nmm.[15]

It has been shown that edge-beveling of orthodontic wires marketed as rectangular can lead to greater play than theoretical values.[11] This was confirmed when stainless steel, nickel-cobalt alloy and beta titanium wires from three manufacturers were compared. Results demonstrated that the average edge bevel contributed to between 0.2-12.9° of the engagement angle. Greater edge bevel lead to greater engagement angle, especially at higher degrees of archwire torsion.[5]

This complex combination of factors affecting torque expression can render clinical decision-making challenging.[7] The application of a torsional moment is considered to be one of the most difficult tasks in orthodontic treatment. A lack of knowledge regarding the appropriate degree of twist required to deliver a pre-determined amount of torsional moment complicates clinical decision-making.[6] In addition, there is inadequate evidence to determine what range of torque moments leads to a corresponding range of clinically applicable alterations in tooth inclination, without causing undue damage to the roots of teeth. The trial and error method of torsional moment application currently practiced can lead to extended treatment times.

The objective of this study is to compare the torque expression of three brands of commercially available metallic self-ligating brackets, two active and one passive, using stainless steel, titanium molybdenum alloy (TMA) and copper nickel titanium (CuNiTi) wire.

4.2 Materials and Methods

An experimental device, the Orthodontic Torque Measurement Device, was utilized to measure torque expression in Nmm. This apparatus consists of a multi-axis force/torque transducer (ATI Industrial Automation Nano 17 Multi-Axis force/torque transducer, Apex, NC, USA) capable of measuring forces and torques in three dimensions while maintaining horizontal and vertical alignment between the bracket and the wire (Figure 4.1). This device isolated and recorded the moment produced by the torsion of the wire segment. Forces and torques in all other planes of space were kept as close to zero as possible.[16]

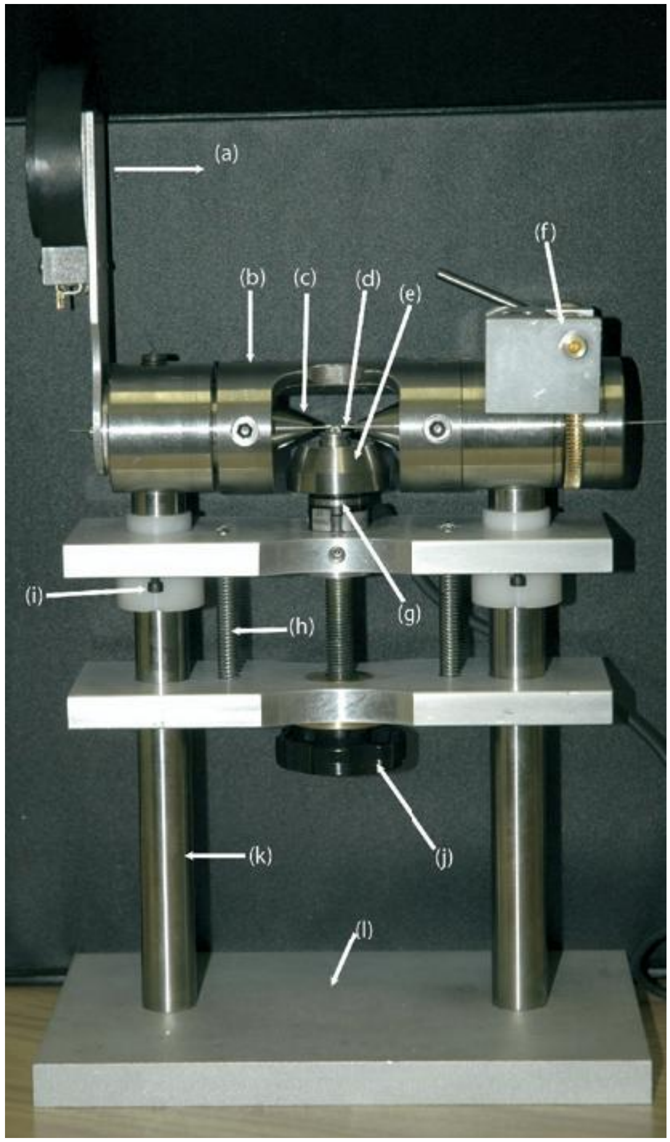

Figure 4.1 Orthodontic Torque Measurement Device

Photograph of the Orthodontic Torque Measurement Device with labeled components:
a, Inclinometer; b, wire support substructure; c, conical wire support and alignment component; d, bracket wire assembly area; e, dual turn table alignment system; f, work gear-based torsion system; g, load cell; h, springs; i, bushings; j, height adjustment system; k, guide rails; l, base support.[16]

The load cell was composed of a silicone strain gauge rated for maximum loads of 25N of transverse force (Fx,Fy), 35N of axial force (Fz) and 0.25Nm moments in all three dimensions. The error of measurement of the force/torque transducer is 1.5% (manufacturer specification). The load cell was used with a data acquisition card which imported the transducer signal to a personal computer (DAC 16-bit E series

NI PCI-6033E, National Instruments, Austin, Tex, USA). LabView data acquisition software (National Instruments, Austin, Tex, USA) was utilized to acquire the signal from the transducer and log the force and torque data in the x dimension to file along with the mean angle of torsion of the wire segment.[16]

The alignment assembly allowed for alignment of the wire and bracket slot and consisted of a base turntable as well as a secondary turntable (Figure 4.2). A digital inclinometer was used to measure the torsion of the wire segment (model T2-7200-1N; US Digital, Vancouver, Wash, USA). The torsion system is worm-gear based, rotating the wire substructure, including the inclinometer, 3° for every full turn of the handle.[16] Torque data was taken every 3° from -21° to 81°.

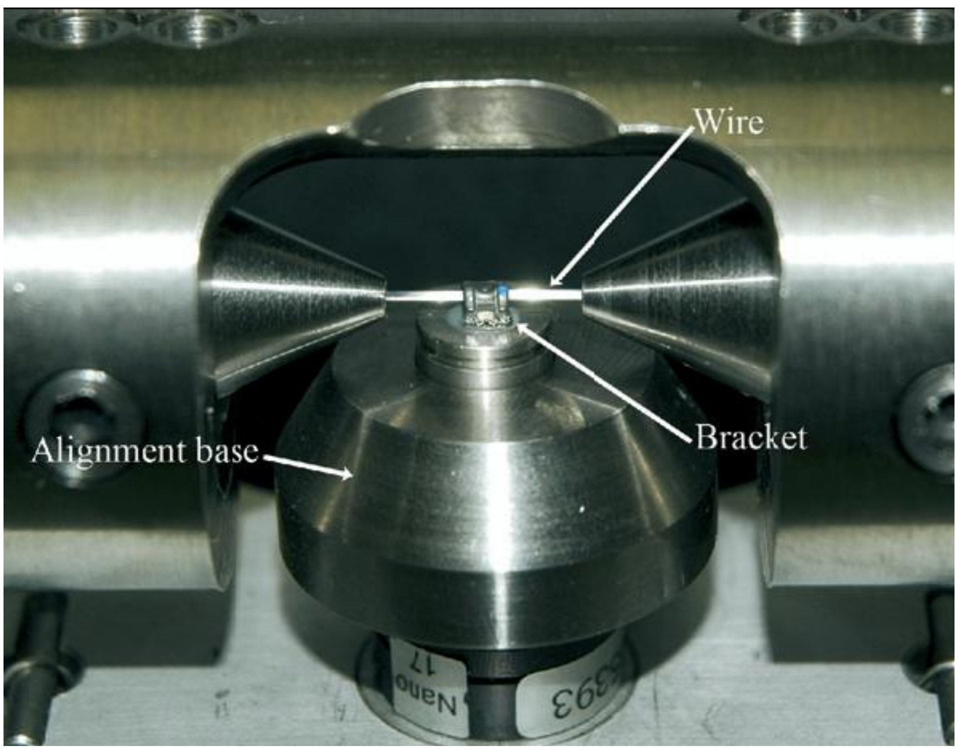

Figure 4.2 Bracket and Wire Alignment in the Orthodontic Torque Measurement Device

One hundred and fifty maxillary right central incisor brackets of each of three types of brackets (total: 450 brackets, In-Ovation-R, GAC, Bohemia, NY, USA; SPEED, Strite Industries, Cambridge, Ontario, Canada; Damon 3MX, Ormco, Orange, CA, USA) were included in the study which measured torque expression using stainless steel, TMA and CuNiTi wire. Each bracket was mounted on a stainless steel cylinder (secondary turntable, figure 4.2) which had been sand-blasted and treated with a degreasing agent (Loctite ODC-Free Cleaner and Degreaser 22355; Henkel, Rocky Hill, Conn, USA). All brackets were bonded to the stainless steel cylinders with an epoxy resin adhesive (Loctite E-20HP Hysol L07KA12771; Henkel, Rocky Hill, Conn, USA) using a bracket mounting jig. The epoxy was allowed to cure for a minimum of 24 hours before testing the brackets. Straight lengths of 0.019x0.025 inch stainless steel orthodontic, titanium molybdenum alloy and copper nickel titanium wire were tested in torsion using a new segment for each experiment (300 series stainless steel lot #07J2J and #05A8A; titanium molybdenum alloy TMA® lot #06E22E and #04B16B; 35° CuNiTi® custom order; Ormco, Glendora, CA, USA). The torque prescription of the brackets did not have an effect on the methods of the study since the true zero torque position was used as a baseline for all brackets. This process involved centering the data cells of each test containing values between -2 and 2 at 0° of wire torsion by shifting the cells (Excel, Microsoft, Redmond, WA, USA).

The Orthodontic Torque Measurement Device[16] was placed in an incubator (Air-shields Isolette® infant incubator, model C100/200-2, series 00, Serial DT 9290, Draeger Medical Systems Inc., Telford, PA, USA) where all tests were conducted at an average temperature of 37.92° (Figure 4.3). All experimental wire segments were also stored in the incubator. All 450 experiments were individually randomized to avoid differences in temperature and calibration of the device from confounding the results. Individual randomization lead to a similar but unequal number of brackets being tested in each bracket-archwire group; however, overall, an equal number of each bracket and wire type were tested. The device was calibrated prior to the start of data collection in May 2008 (ATI Industrial Automation, Apex, NC, USA). Data was collected during an average of two sessions per week for 14 weeks.

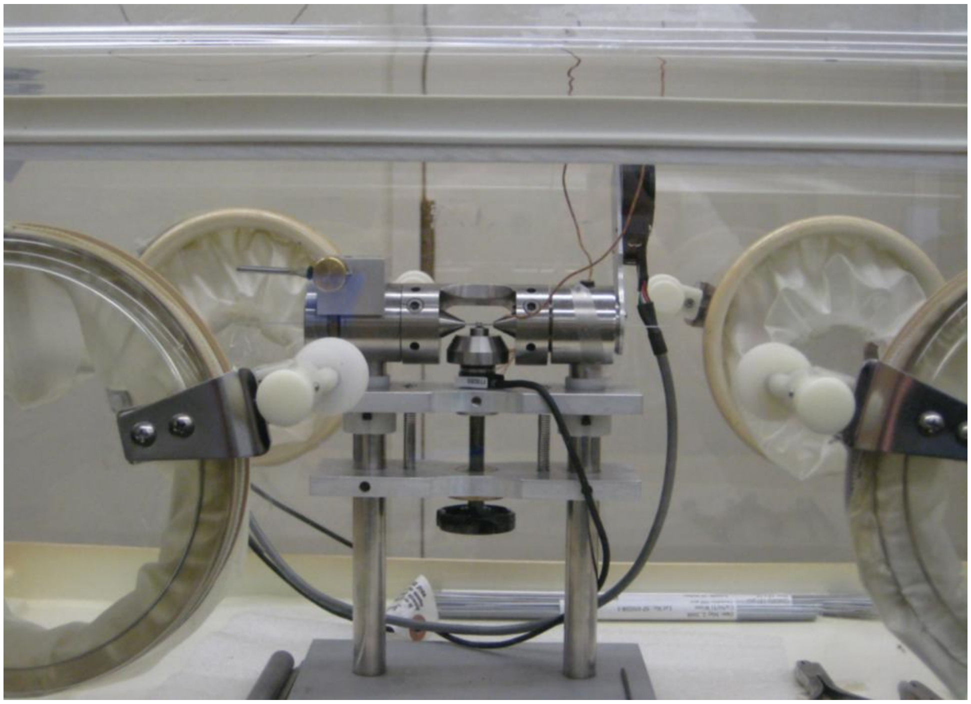

Figure 4.3 Orthodontic Torque Measurement Device Within Incubator

A potential cause of variability in torque expression is variation in bracket slot dimensions.[7] Due to the high variability in torque expression that was observed in this study, a sample of 10 of each bracket type (In-Ovation-R, SPEED and Damon 3MX) were measured for width at the base of the bracket and at the superior aspect of the bracket slot. The superior aspect of the active self-ligating brackets was measured at the point where the clip closed on the slot wall. For the Damon 3MX bracket, the superior aspect of the slot was measured at a 90° angle to the sliding door, on the slot wall. Due to the small sample size, statistical analysis was not performed; however, descriptive statistics highlight the variability in bracket slot shapes and sizes. Measurements were made with an optical comparator which projected and magnified the image of each bracket twenty times (Mitutoyo PH 3500 Profile Projector, lens resolution: 0.0001 inch, comparator resolution: 0.000004 inch, Japan). Crosshairs on the projection screen were used to measure the bracket slot width. Error measurements were taken for each bracket and were found to be ≤0.0001inch.

In addition to bracket slot width, variation in wire dimensions may lead to variation in torque expression measurements.[8-10] Ten straight length sections of each 35°CuNiTi, TMA and stainless steel wire were measured for width and height in three locations along each wire with a micrometer (Mitutoyo IP 54, Japan, accuracy within 0.00005 inch). Descriptive statistics are presented in the results section.

4.3 Results

The sample size for this study was based on calculations from a previous study using the Orthodontic Torque Measurement Device.[16] The calculation was based on the ability to detect a difference of ≥5Nmm (minimum value for clinical significance),[15] with α=0.05 and β=0.1. An average sample size of 50 of each bracket type was determined to be adequate based on calculations for angles 12° (81), 24° (40), 36° (19) and 48° (54). Error analysis was not conducted for this study; however, it is expected to compare to the error measurements of Badawi et al who demonstrated a coefficient of variation (standard deviation/mean) of 0.45-1.27% for a torsional angle of 48° and 1.63-7.86% for a torsional angle of 36°, with an average coefficient of variation of 2.7%.[16] Several brackets, particularly SPEED, submitted to torsion with stainless steel wire de-bonded at torsional angles over 42°. This led to a decrease in the sample size for statistical calculations at 48° of wire torsion, however, this degree of wire torsion produced moments above the clinically applicable range of 5-20Nmm.[15]

At 12°, the data approached normal distribution; however, at 24°, 36° and 48°, the distribution of the data was not normal. As the angle of torsion increased, the data deviated farther from the expected values for a normal distribution (Figures A-D, Appendix C). Levene's Test of Equality of Error Variances indicated that the equal variance assumption was violated at all four angles of torsion. Figure 4.4 demonstrates the trend towards increasing variance with increasing torsional angle for all wire types combined. Similar results were reported for stainless steel wire by Badawi et al.[16] The variance of each bracket and wire combination at 24° of wire torsion is depicted in Figure 4.5. The variance of all three brackets with stainless steel wire is greater than with TMA and CuNiTi wire. This is visible in Figure 4.5 and may be due to the generation of higher torque moments with stainless steel wire, leading to a larger coefficient of variation (SD/mean). Box's Test of Equality of Covariance Matrices indicated a lack of equality of covariance matrices for all three bracket types. Figure 4.6 displays the mean torque expression of each bracket and wire combination. The importance of wire alloy as a factor is highlighted since the average plots cluster at three levels of torque expression based on wire alloy. Differences between brackets are also evident.

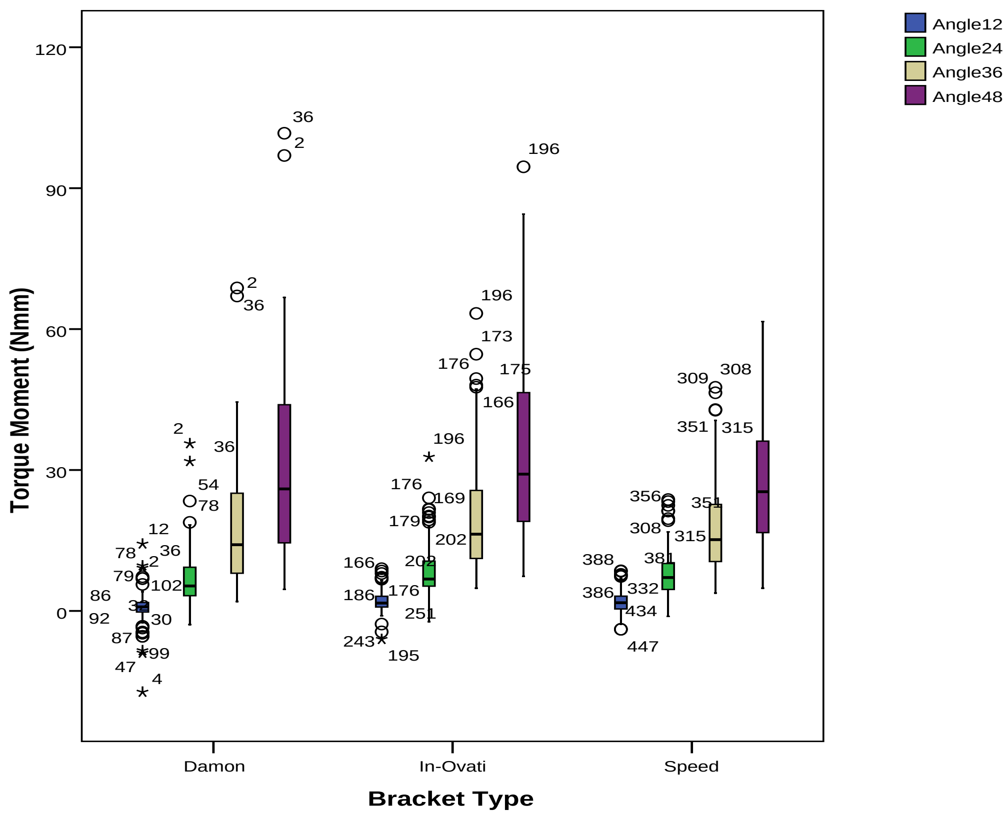

Figure 4.4 Median Torque Expression According to Torsional Angle

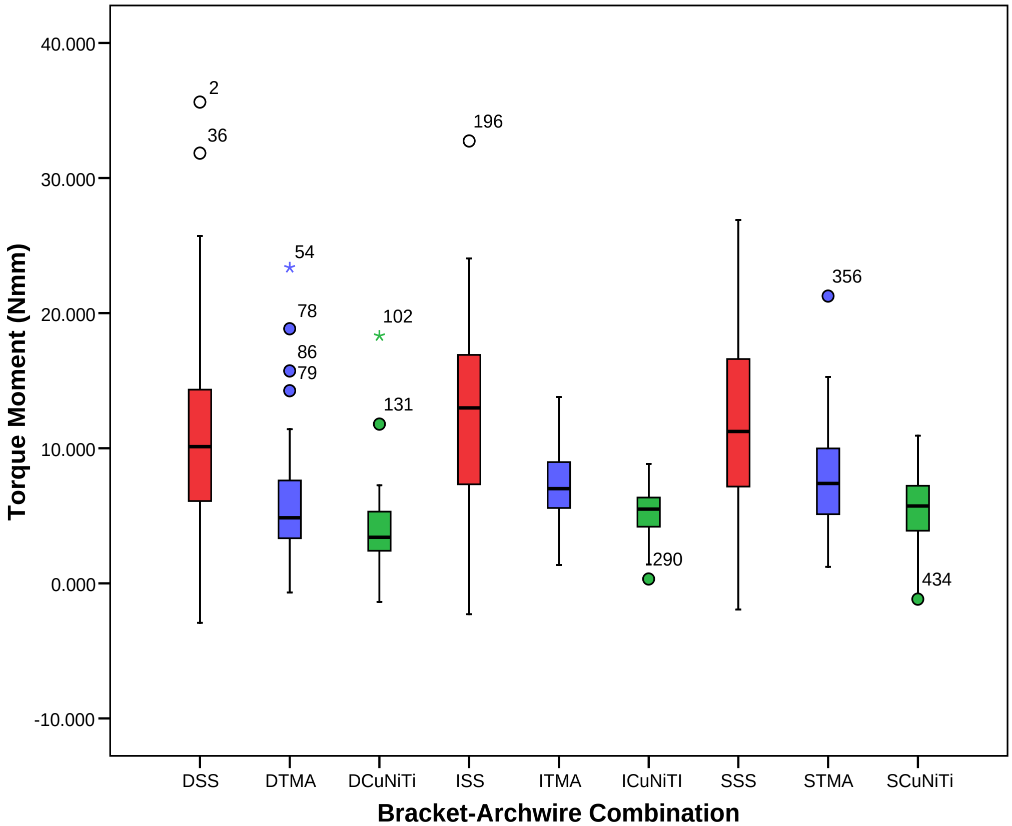

Figure 4.5 Median Torque Expression for All Bracket-Wire Combinations at 24° of Wire Torsion

D=Damon 3MX
I=In-Ovation-R
S=SPEED
SS=Stainless Steel
TMA=Titanium Molybdenum Alloy
CuNiTi=Copper Nickel-Titanium

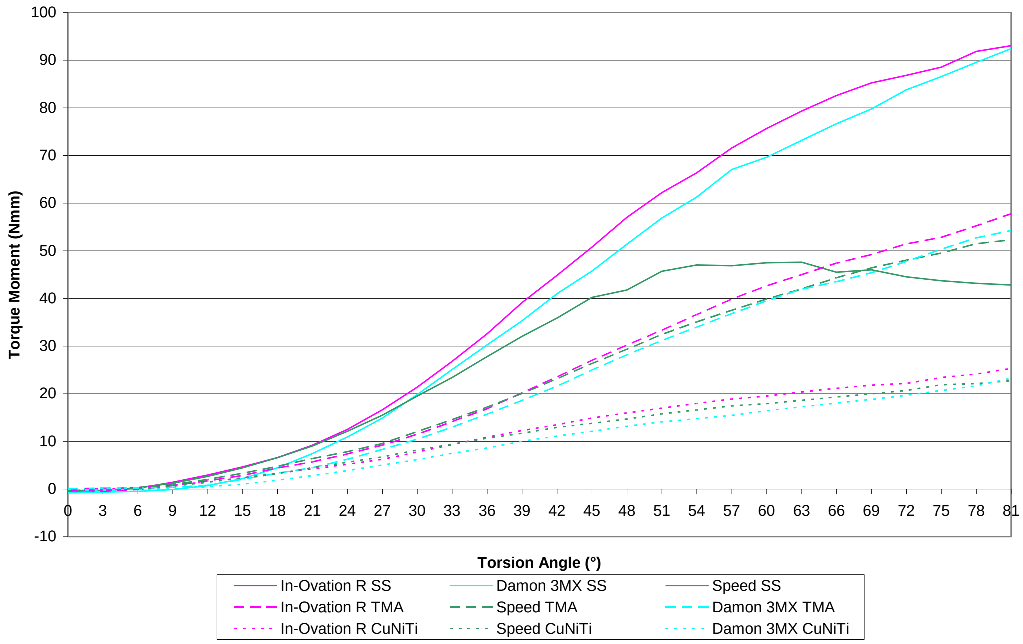

Figure 4.6 Average Torque Expression

Multivariate analysis of variance (MANOVA) and one way analysis of variance (ANOVA) with specific contrast tests were conducted with a statistical package (SPSS, Chicago, Ill, USA) to identify any significant differences in torque expression between brackets. Descriptive statistics are presented for all four angles (Tables 4.1-4.4). A high degree of variance in torque measurements was observed within each bracket type as evidenced by high standard deviation values. This high degree of variation may also be observed in the graphs of each set of experiments grouped by bracket and wire combination featured in Figures E-M in Appendix D.

Table 4.1 Descriptive Statistics for Torsional Angle 12°

Wire alloy	Bracket	Mean Torque (Nmm)	Standard Deviation (Nmm)	N
	Damon	0.687	4.740	52
SS	In-OvationR	2.899	3.094	50
	SPEED	2.655	2.654	50
	Damon	0.770	2.955	49
TMA	In-OvationR	1.717	1.813	50
	SPEED	1.966	2.501	49
	Damon	0.471	1.826	52
CuNiTi	In-OvationR	1.492	1.119	49
	SPEED	1.398	1.888	49

Table 4.2 Descriptive Statistics for Torsional Angle 24°

Wire alloy	Bracket	Mean Torque (Nmm)	Standard Deviation (Nmm)	N
	Damon	10.899	7.274	52
SS	In-OvationR	12.493	6.635	50
	SPEED	12.102	6.405	50
	Damon	6.232	4.647	49
TMA	In-OvationR	7.354	2.659	50
	SPEED	7.833	4.023	49
	Damon	3.865	3.098	52
CuNiTi	In-OvationR	5.185	1.832	49
	SPEED	5.592	2.614	49

Table 4.3 Descriptive Statistics for Torsional Angle 36°

Wire alloy	Bracket	Mean Torque (Nmm)	Standard Deviation (Nmm)	N
	Damon	30.236	11.492	52
SS	In-OvationR	33.555	11.141	50
	SPEED	27.800	9.050	49
	Damon	15.669	7.470	49
TMA	In-OvationR	16.826	5.541	50
	SPEED	17.172	6.243	49
	Damon	8.611	5.759	52
CuNiTi	In-OvationR	10.838	3.644	49
	SPEED	10.669	3.636	49

Table 4.4 Descriptive Statistics for Torsional Angle 48°

Wire alloy	Bracket	Mean Torque (Nmm)	Standard Deviation (Nmm)	N
SS	Damon	51.356	13.543	47
	In-OvationR	57.002	14.735	46
	SPEED	41.732	9.901	34
TMA	Damon	28.131	10.218	49
	In-OvationR	30.147	7.137	50
	SPEED	29.313	7.866	49
CuNiTi	Damon	13.152	6.578	52
	In-OvationR	15.944	5.325	49
	SPEED	14.676	4.659	49

Torque moments in the range of 5-20Nmm have been proposed as clinically effective.[15] This range of torsional moments can be achieved with the following range of torsional angles which differs depending on wire alloy.

With stainless steel wire (Figure 4.7), torque moments of 5Nmm were observed at approximately 15°of archwire torsion in the active self-ligating brackets tested, In-Ovation-R and SPEED; whereas this moment was reached at approximately 18° for the passive self-ligating bracket, Damon 3MX. In-Ovation-R reached a moment of approximately 20Nmm at 30° of torsion, with SPEED and Damon 3MX expressing slightly less torque. The range of clinically applicable torsional angles is approximately 15° for SPEED and In-Ovation-R and 12° for Damon 3MX (5-20Nmm).

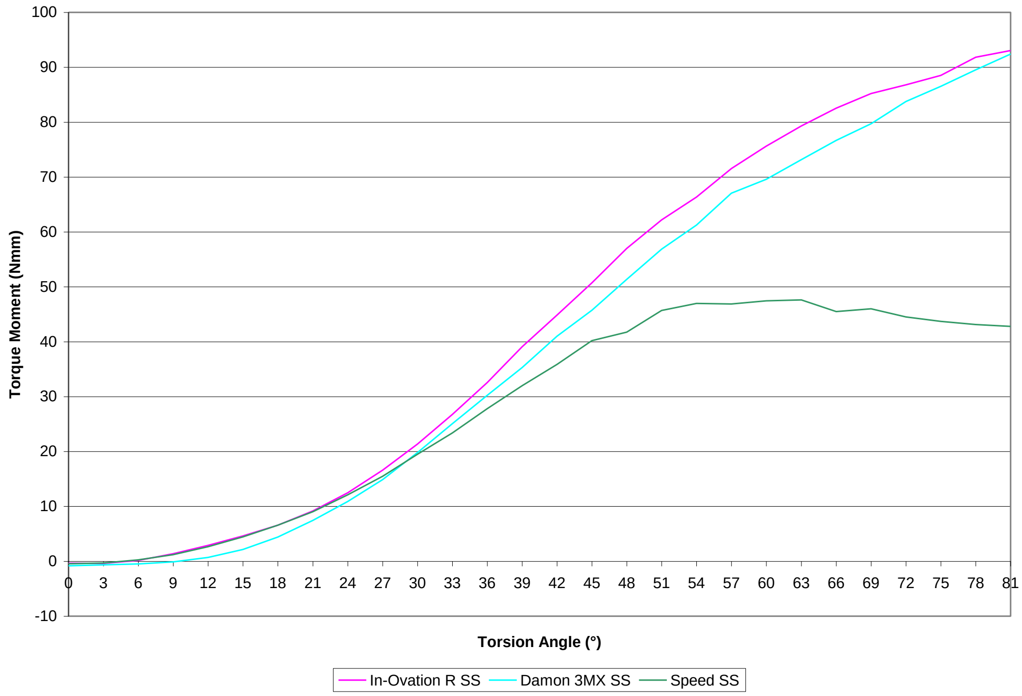

Figure 4.7 Average Torque Expression for Stainless Steel Wire

For TMA wire (Figure 4.8), the clinically applicable range of wire torsion was greater than for stainless steel wire. For the SPEED bracket, approximately 5Nmm of torsional moment was attained at approximately 18° of wire torsion. At 39°, approximately 20Nmm of torque moment was produced for a range of 21°. Damon 3MX and In-Ovation-R only began to express clinically effective torque moments at approximately 21°. The Damon 3MX bracket reached the maximum for a clinically valid torque moment at approximately 42 degrees for a range of 21°; whereas, the In-Ovation-R bracket reached the same point at approximately 39° for a range of 18°.

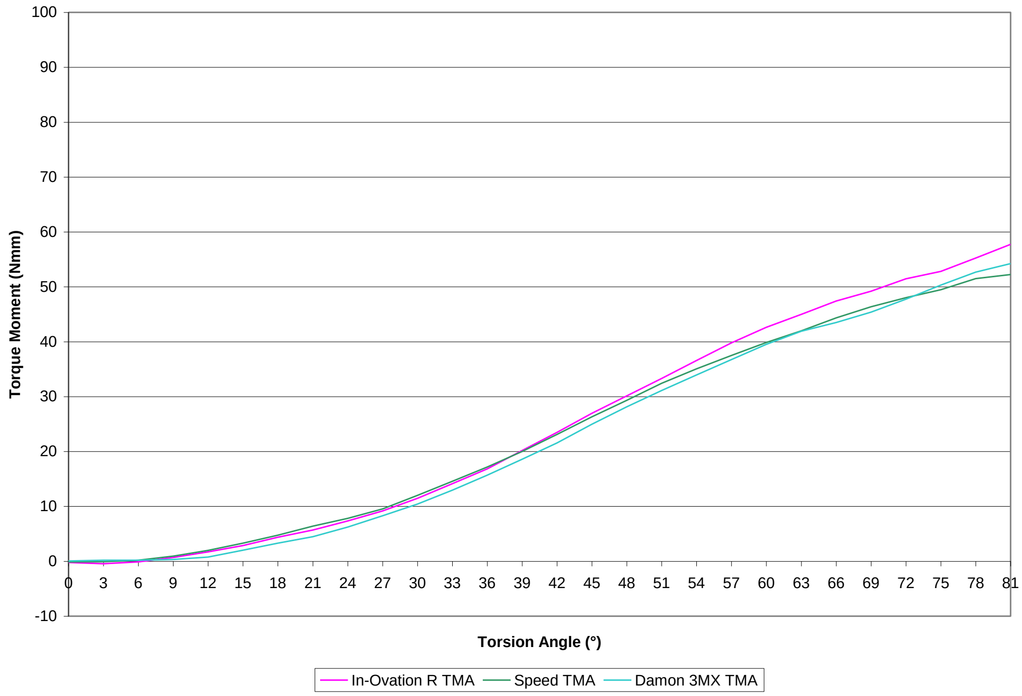

Figure 4.8 Average Torque Expression for TMA Wire

The CuNiTi wire (Figure 4-9) had a far greater clinically applicable range of wire torsion than both stainless steel and TMA; from approximately 21-72° for SPEED (51°), 27-78° for Damon 3MX (51°) and 24-63° for In-Ovation-R (39°).

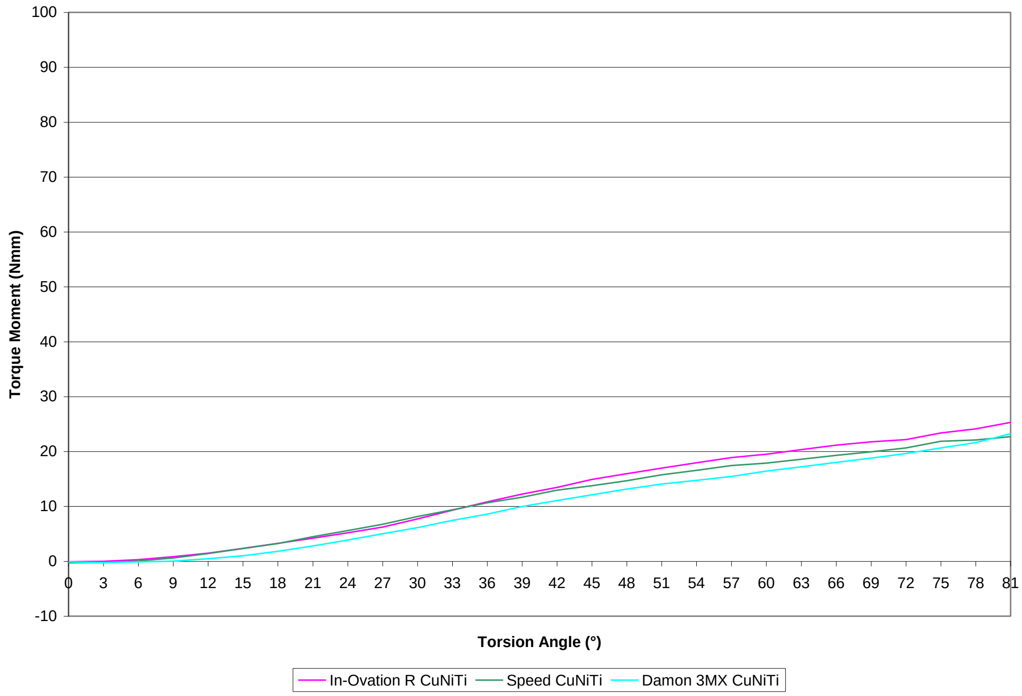

Figure 4.9 Average Torque Expression for CuNiTi Wire

Statistical comparisons between bracket types were made at 12°, 24°, 36° and 48° and are presented in Table 4.5. Equal variance was not assumed due to the violation of this model assumption. Significance values take unequal variance into account.

Table 4.5 ANOVA with Contrast Tests: Statistically Significant Differences

Torsional Angle	Wire Alloy	Bracket Comparisons	Mean Torque Difference (Nmm)	Significance
12°	SS	I-D	2.213	0.006
		S-D	1.968	0.019
	TMA	S-D	1.195	0.033
	CuNiTi	I-D	1.021	0.001
		S-D	0.927	0.014
24°	SS	NS	NS	NS
	TMA	NS	NS	NS
	CuNiTi	I-D	1.320	0.010
		S-D	1.727	0.003
36°	SS	I-S	4.755	0.022
	TMA	NS	NS	NS
	CuNiTi	I-D	2.227	0.022
		S-D	2.057	0.034
48°	SS	D-S	9.625	<0.001
		I-S	15.270	<0.001
	TMA	NS	NS	NS
	CuNiTi	I-D	2.792	0.021

D=Damon 3MX
I=In-Ovation-R
S=SPEED
NS: No significant difference

At 12° of wire torsion with stainless steel wire, there was a small but statistically significant difference in torque expression between the Damon 3MX bracket and the In-Ovation-R and SPEED brackets. The In-Ovation-R and SPEED brackets exhibited torque moments approximately 2Nmm greater than the Damon 3MX bracket. With TMA wire, SPEED exerted approximately a 1Nmm greater torsional moment than Damon 3MX. Finally, with the copper NiTi wire, In-Ovation-R and SPEED both expressed approximately 1Nmm more torque than the Damon 3MX bracket. Thus, the Damon 3MX bracket expressed the least torque with all wire alloys at 12° of wire torsion.

At 24° of wire torsion with stainless steel wire, there were no statistically significant differences in torque expression between bracket types though the difference in torque expression between Damon 3MX and the other two brands was in the 1.2-1.5Nmm range, with Damon 3MX registering lower moments. With TMA wire, SPEED expressed 1.6Nmm more torque than Damon 3MX. When testing CuNiTi wire, In-Ovation-R expressed 1.3Nmm and SPEED expressed 1.7Nmm more torque than the Damon 3MX bracket.

At 36° of wire torsion with stainless steel wire, a change in the order of brackets according to torque expression was noted. In-Ovation-R demonstrated the greatest torque moment, followed by Damon 3MX and then by SPEED. Only the difference between the highest, In-Ovation-R and the lowest, SPEED was found to be statistically significant with a difference of almost 5Nmm. There were no statistically significant differences between brackets with TMA wire. In contrast, with the CuNiTi wire, In-Ovation-R and SPEED both expressed approximately 2Nmm more torque than Damon 3MX.

At 48° of wire torsion, all three brackets showed statistically significant differences when compared to the others with stainless steel wire. In-Ovation-R produced the highest torsional moment, followed by Damon 3MX and finally by SPEED. In-Ovation-R exerted a 15.3Nmm higher torque moment than SPEED and a 5.6Nmm higher torque moment than the Damon 3MX bracket. Damon 3MX applied 9.6Nmm more torsional moment than SPEED. Thus, the SPEED bracket exerted significantly less torsional moment than both In-Ovation-R and Damon 3MX. This drop in torque expression of SPEED compared to the other two brackets can be seen in Figures 4.6 and 4.7. Comparisons between brackets with TMA wire were not statistically significant. With CuNiTi wire, In-Ovation-R expressed approximately 3Nmm more torque than Damon 3MX.

Boxplots of the width of the bracket base and superior aspect "top" of the bracket slot are depicted in Figures 4.10 and 4.11. The SPEED bracket displays a greater median width at the base and a smaller median width at the top than the other two brands of brackets. The SPEED brackets also displayed the least variance as evidenced by the overall size of the boxplots in Figure 4.12 in comparison to the In-

Ovation-R and Damon 3MX brackets. The difference between the median widths at the base and top of the SPEED bracket slot are also apparent in Figure 4.12.

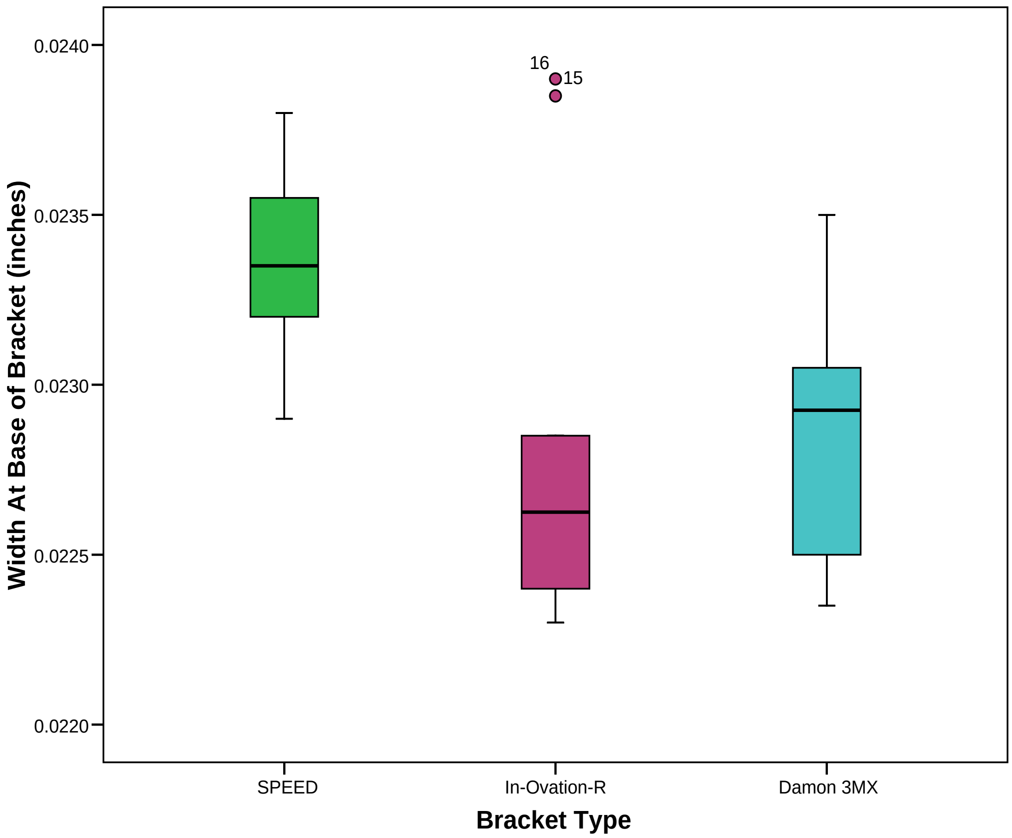

Figure 4.10 Median Bracket Base Width

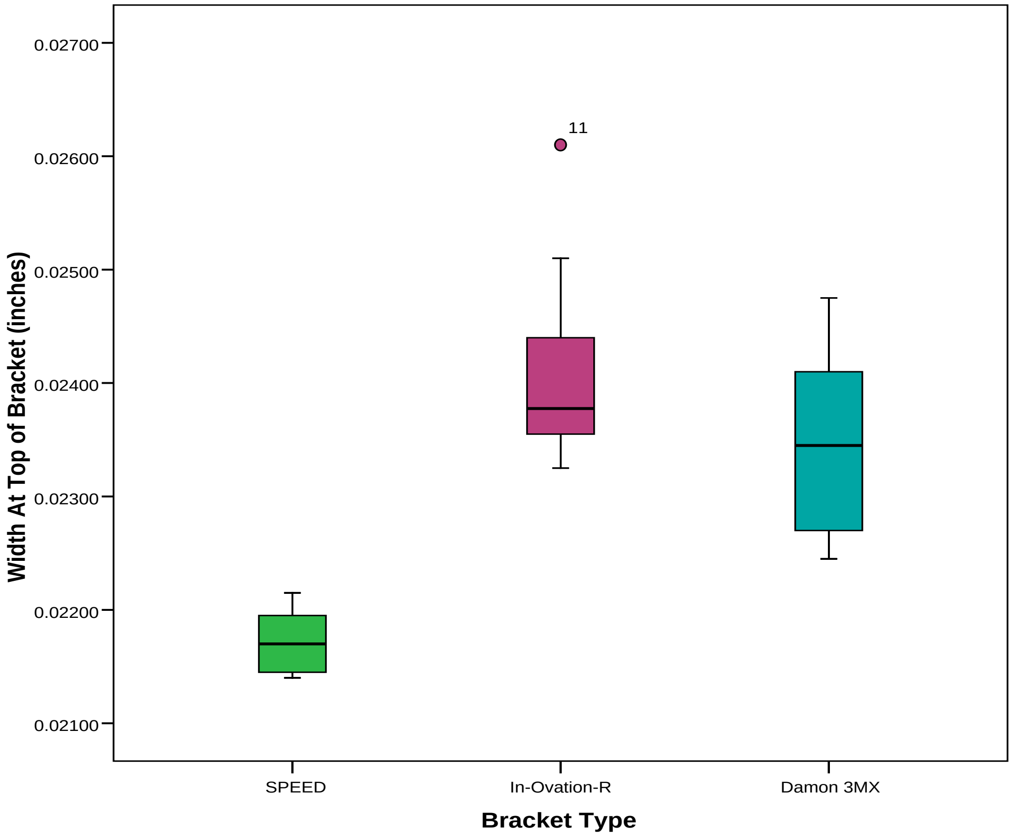

Figure 4.11 Median Bracket Top Width

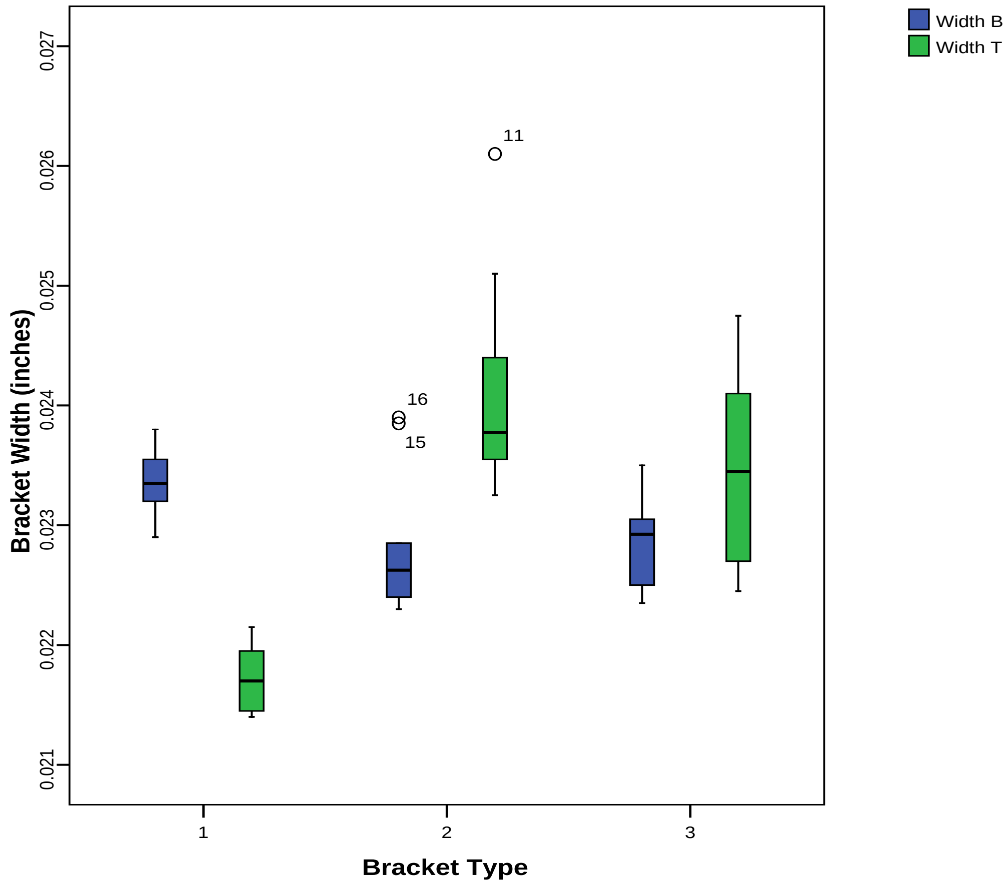

Figure 4.12 A Comparison of Bracket Base and Top Width

The measurements of the bracket width at the base and top of the slot are presented in Tables 4.6 and 4.7. These values echo the visual representation of the bracket widths in the boxplots. As previously stated, the SPEED bracket is wider at the base and narrower at the top than the In-Ovation-R and Damon 3MX brackets. The range and standard deviations for the SPEED brackets are lower than other brands.

Table 4.6 Bracket Base Width Measurements

Bracket	Mean Width at Base (inches)	Standard Deviation	Range
SPEED	0.023365	0.0002769	0.0009
In-Ovation-R	0.022840	0.0005748	0.0016
Damon 3MX	0.022875	0.0004211	0.0012

Table 4.7 Bracket Top Width Measurements

Bracket	Mean Width at Top (inches)	Standard Deviation	Range
SPEED	0.0217300	0.0002689	0.00075
In-Ovation-R	0.0241000	0.0008778	0.00285
Damon 3MX	0.0235000	0.0007835	0.00230

Another potential cause for variation in torque expression is variation in wire dimension.[8-10] In order to explore this possibility, a sample of 10 straight lengths of each wire type (Ormco stainless steel, TMA and CuNiTi) were measured for height and width in three locations along each wire. Results are presented in tables 4.8 and 4.9. The CuNiTi wires were undersized in height; whereas, the TMA wires were undersized in width. The stainless steel wire showed lower standard deviations and ranges than the other two alloys for both height and width.

Table 4.8 Wire Height Measurements

Wire alloy	Mean1 (inches)	SD1	Range1	Mean2 (inches)	SD2	Range2	Mean3 (inches)	SD3	Range3
SS	0.0250450	0.0001165	0.00040	0.0250650	0.0001292	0.00050	0.025045	0.0000725	0.0003
TMA	0.0248250	0.0000716	0.00020	0.0248200	0.0000752	0.00025	0.025010	0.0004960	0.0017
CuNiTi	0.0243700	0.0000537	0.00020	0.0246050	0.0006885	0.00230	0.024790	0.0008928	0.0024

Table 4.9 Wire Width Measurements

Wire alloy	Mean1 (inches)	SD1	Range1	Mean2 (inches)	SD2	Range2	Mean3 (inches)	SD3	Range3
SS	0.0193300	0.0001110	0.00030	0.019320	0.0000675	0.0002	0.0193050	0.0000437	0.00010
TMA	0.0189100	0.0000614	0.00020	0.018930	0.0001457	0.0005	0.0189450	0.0001012	0.00035
CuNiTi	0.0194400	0.0001264	0.00050	0.019420	0.0000856	0.0003	0.0194800	0.0001636	0.00060

4.4 Discussion

This study was an in-vitro study designed to compare the torque capacity of three brands of commercially available self-ligating brackets. Figure 4.7 shows the climb in torsional moment with respect to stainless steel wire torsion. It is evident that torque expression begins earlier for the active self-ligating brackets, In-Ovation-R and SPEED. These brackets begin to express torque at approximately 7.5°; whereas, the Damon 3MX bracket begins to express torque at approximately 12°. Differences in bracket slot design likely account for this disparity. These engagement angles (play) are similar to results from Badawi et al, who noted a 7.5° engagement angle for In-Ovation-R and SPEED and 15° for the Damon 2 bracket.[16] Both of these studies demonstrate smaller engagement angles for the active SL designs than previously reported for conventional brackets (10.9-23.7°).[5]

Variations in bracket slot dimensions have been proposed as a cause of variability in torque expression.[5,10,17] From Figure 4.13, one can see that the slot shapes of the Damon 3MX (a), In-Ovation-R (b) and SPEED (c) brackets vary. A visible divergence of the slot walls from the base to the top of the bracket slot is seen in the Damon 3MX and In-Ovation brackets. In contrast, the SPEED bracket is convergent from the base of the slot to the top, displaying a distinct keyhole shape, with rounded corners where the concave base meets the walls. These shape differences which can be seen visually in Figure 4.13 are confirmed by the width measurements presented in the results section (Tables 4.8 and 4.9).

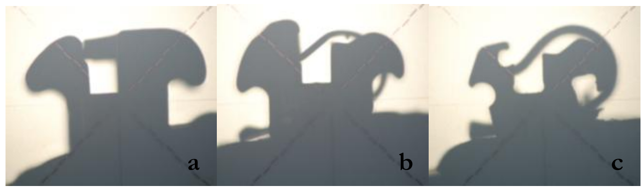

Figure 4.13 Photographic Images of Bracket Profiles

a, Damon 3MX Bracket; b, In-Ovation-R Bracket; c, SPEED Bracket

In 2004, Cash et al evaluated the slot size of several brands of 0.022 inch slot orthodontic brackets. The Damon II SL bracket was found to have convergent slot walls with its base oversized by 17%.[18] These results differ from the results of this study measuring the Damon 3 MX bracket. Differences in the slot design of the Damon 3MX vs the Damon 2 brackets may be the cause. Meling et al examined bracket slot height in Ormco medium twin, standard edgewise 0.018 inch brackets and found that the mean height was outside of the range commonly quoted for this bracket type.[17] The results of a small sample of brackets measured in this thesis confirm this statement since all three brackets are marketed as having a slot width of 0.022 inches; however, all measurements, with the exception of the top of the SPEED bracket, surpassed this value. These results must be interpreted with caution since the sample size was small and the caliper was zeroed but not calibrated prior to taking the measurements. The theoretical play (engagement angle) can be calculated based on equations presented by Meling et al[17] using archwire dimensions and bracket slot width. With stainless steel wire, play should theoretically be 9.97° for Speed, 8.58° for In-Ovation-R and 8.67° for Damon 3 MX. With TMA wire, the theoretical play should be 11.10° for Speed, 9.67° for In-Ovation-R and 9.77° for Damon 3MX. Finally, for CuNiTi wire, play should be 7.49° for Speed, 6.44° for In-Ovation-R and 6.51° for Damon 3MX. Since the results demonstrate that the actual play for Speed and In-Ovation-R is approximately 7.5° whereas, it is approximately 12° for Damon 3MX, factors such as bracket height and the presence or absence of an active clip may contribute to play.

Table 4.10 demonstrates the relative contribution of the variables bracket and wire to torque expression as well as their interaction. It is clear that the wire alloy is a far more important determinant of torque expression than the bracket type as it accounts for 56.5% of the variation. The bracket type accounts for 13.3% of variation in torque expression results. Finally, the combined effect of bracket and wire (interaction) is responsible for 5.7% of the variance in torque expression. This means that a bracket will behave differently depending on the wire alloy that is used with it. Likewise, the performance of a wire will be altered by the bracket selected. The total variation explained by the bracket type, wire alloy and their combination (interaction) is 75.5%. The remaining 25% may be attributed to the following variables; wire edge beveling, wire dimension variation, bracket dimension variation and forces in other dimensions.

Table 4.10 MANOVA: Multivariate Tests

Variable	P-value	% Variance Explained by Variable
Bracket	<0.0001	13.3
Wire	<0.0001	56.5
Bracket*Wire	<0.0001	5.7

Profile plots (Figures 4.14-4.17) highlight the effect of wire alloy on torque expression. They also demonstrate an interaction of the bracket and wire type producing a combined effect on torque expression.

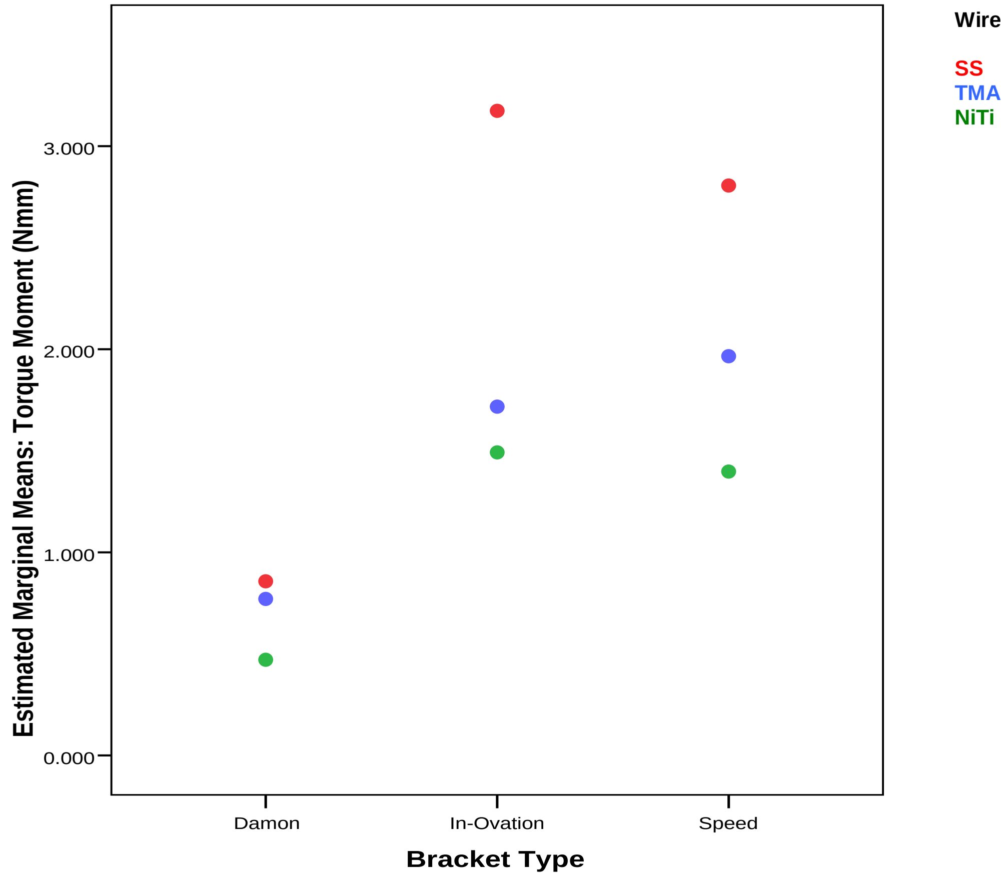

Figure 4.14 Profile Plot at 12° of Wire Torsion

At 12° of wire torsion (twist), wire alloy is not as important a factor in torque expression in the Damon 3MX bracket as it is in the SPEED and In-Ovation-R brackets which both show significantly higher torque expression with stainless steel wire (Figure 4.14). In addition, the Damon 3MX bracket expresses significantly less torque than the other two brands at 12° of archwire torsion, regardless of the wire type. This may be due to its slot design.

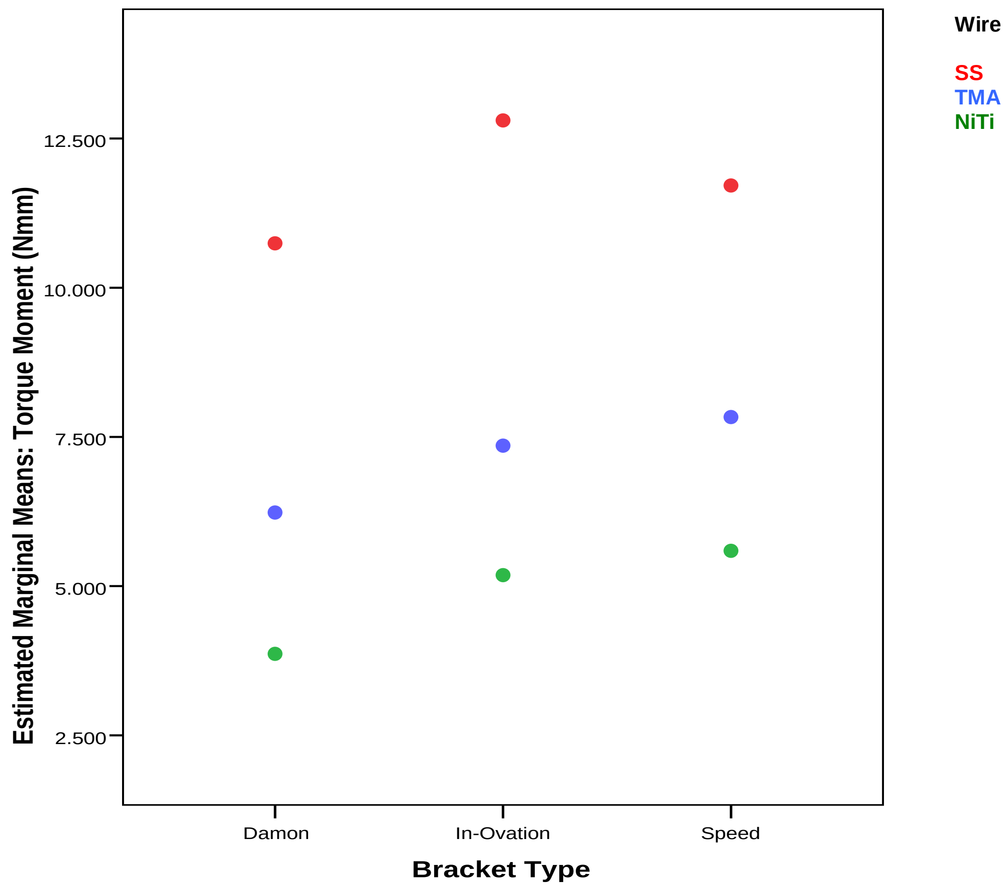

Figure 4.15 Profile Plot at 24° of Wire Torsion

At 24° of wire torsion (twist), wire alloy becomes an equally important factor in determining torque expression for all three brackets; however, there are differences in torque expression between bracket types (Figure 4.15). In-Ovation-R expresses the most torque, followed by SPEED and then by Damon 3MX.

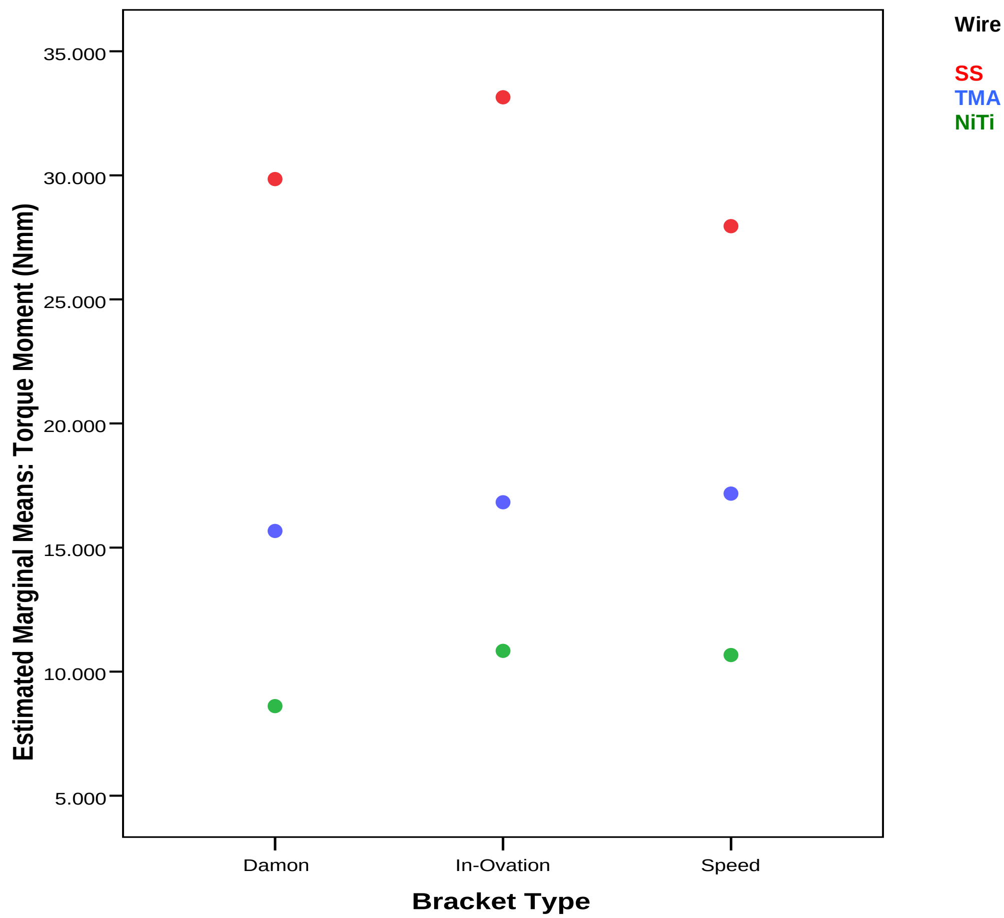

Figure 4.16 Profile Plot at 36° of Wire Torsion

At 36° of wire torsion (twist), the difference between the torque expression with stainless steel and the other two alloys is heightened with stainless steel expressing greater torque moments regardless of bracket type (Figure 4.16). There are differences between bracket types with In-Ovation-R expressing the highest torque moments. With stainless steel wire, the bracket order has switched compared to data at 24° of wire torsion. Now Damon 3MX expresses more torque than SPEED.

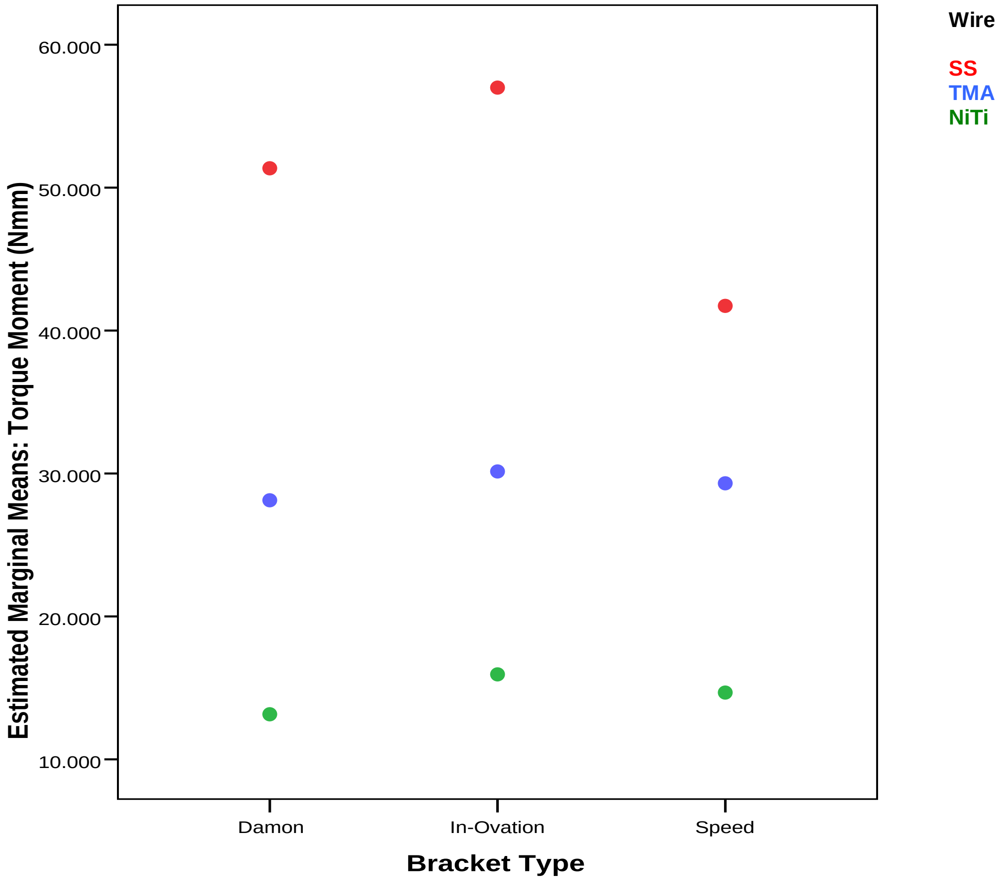

Figure 4.17 Profile Plot at 48° of Wire Torsion

Forty-eight degrees of wire torsion (twist) is generally considered above the amount of torsion that is incorporated into a stainless steel wire clinically. Moments above 50Nmm (twice the maximum recommended level) were observed in the In-Ovation-R and Damon 3MX brackets. These torque moments were roughly 1.5-2 times the moments generated with TMA wire (Figure 4.17). The SPEED bracket behaved differently with a moment of approximately 40Nmm at 48° with stainless steel wire compared to over 50Nmm with the other brands. The climb in torque moment for the SPEED bracket was approximately the same between CuNiTi, TMA and stainless steel wire unlike the other two brands which spiked when torque was applied with stainless steel wire.

It is interesting to note a relative lack of statistically significant differences between bracket brands with stainless steel wire at 24° and 36°, except for a statistically and clinically significant difference of almost 5Nmm between In-Ovation and SPEED at 36°, with In-Ovation-R expressing more torque. This may be due to distortion of the SPEED bracket. Though statistically significant differences were found between all bracket types, the mean torsional moments produced at 48° of torsion with stainless steel wire were 41.7-57.0Nmm, which are considered to be excessive for clinical application. The SPEED bracket did not produce torque moments that increased at the same rate as the Damon 3MX and In-Ovation-R brackets when the torsional angle increased beyond approximately 42° (Figure 4.7). This finding may be due to bracket deformation. The trend of the SPEED bracket to fall below the torque moments produced by the Damon 3MX and In-Ovation-R brackets at high torsional angles (Figure 4.7) is in contrast to its performance at 12°.

4.5 Conclusions

The objective of this study was to identify differences in torque expression between three brands of metallic self-ligating brackets, two active and one passive, with stainless steel, TMA and CuNiTi wire. Our conclusions are the following:

- Torque moments of 5Nmm were observed at approximately 15°of stainless steel archwire torsion in the active self-ligating brackets tested, In-Ovation-R and SPEED; whereas this moment was reached at approximately 18° for the passive self-ligating bracket, Damon 3MX. This implies that torque expression with stainless steel wire can only begin to occur if archwire torsion or bracket prescription is at least 15° in the active SL brackets and 18° in the Damon 3MX bracket. All three brackets reached a moment of approximately 20 Nmm at 30° of torsion. The range of clinically effective torque was approximately 15° for the active SL brackets and 12° for the Damon 3MX bracket.

- For TMA wire, the range of clinically effective torque moments was greater than for stainless steel wire. The SPEED bracket attained approximately 5Nmm of torsional moment at 18° of wire torsion. At 39° approximately 20Nmm of torque moment was produced. The Damon 3MX and In-Ovation-R brackets only began to express clinically effective torque moments at approximately 21°. The Damon 3MX bracket reached the maximum for a clinically valid torque moment at approximately 42 degrees; whereas, the In-Ovation-R reached the same point at approximately 39°. The range of clinically effective torque was approximately 21° for the SPEED and Damon 3MX brackets and 18° for the In-Ovation-R bracket.
- The CuNiTi wire had an extremely large clinically applicable range of wire torsion; from approximately 21-72° for SPEED, 27-78° for Damon 3MX and 24-63° for In-Ovation-R. The range of clinically effective torque was approximately 51° for the SPEED and Damon 3MX brackets and 39° for the In-Ovation-R bracket. These extremely wide ranges of clinically applicable torsional angles are likely to make clinical torque application with CuNiTi wire unpredictable.
- Torque expression begins earlier for the active self-ligating brackets, In-Ovation-R and SPEED. These brackets begin to express torque at approximately 7.5° of wire torsion; whereas, the Damon 3MX bracket begins to express torque at approximately 12° of wire torsion.

4.6 References

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Chapter 5. General Discussion and Conclusions

Torque expression poses a daily challenge in the treatment of malocclusions. The application of a torsional moment is considered to be one of the most difficult tasks in orthodontic treatment. A lack of knowledge regarding the appropriate degree of archwire twist required to deliver a pre-determined amount of torsional moment complicates clinical decision-making.[1]

Stainless steel has traditionally been the wire alloy of choice for torque application; however, β-titanium and nickel titanium or copper nickel titanium are also currently being employed for this purpose. Though, there is no evidence regarding the degree of wire twist that is required to produce a clinically effective range of torsional moments for these wire alloys.

This thesis was designed to examine torque expression *in vitro* using stainless steel, TMA and CuNiTi wire. Results indicate earlier torque expression with SPEED and In-Ovation-R brackets than with the Damon 3 MX bracket. This difference may be due to differences in bracket slot shape and bracket design. This pattern continues until approximately 30° of wire torsion, when the SPEED bracket slows its increase in torsional moment relative to twist angle. Bracket deformation may account for this occurrence. Differences in torque expression between brackets were present; however, wire alloy was a far more important determinant of torque expression. Increased torsional stiffness lead to increased torque expression with stainless steel expressing the highest torsional moments, followed by β-titanium and then by CuNiTi.

5.1 Strengths of the Study

This study employed a sophisticated device capable of isolating the moment of the couple in the x-dimension with a multi-axis force-torque transducer previously described.[2] A large sample size of brackets and wires were tested under temperature controlled conditions. Due to the heat-sensitivity of CuNiTi wire, temperature control in an incubator was necessary throughout the 14 weeks of data collection. Wires were stored inside the incubator to mimic intra-oral temperature due to the importance of temperature on the elastic properties of thermoelastic nickel titanium archwire alloys. It is necessary to test under experimental conditions reproducing the intra-oral environment.[3] Once the wire was secured in the bracket slot and the door was closed, data was collected every three degrees by logging the data to file on a computer. A new wire and bracket were used for each test in order to prevent deformation which would lead to a reduction in torque expression.[4] All wire and bracket combinations were randomized individually in order to prevent factors such as loss of calibration or temperature from confounding the results.

5.2 Limitations of the Study

Though the sample size was relatively large for this type of study, the study was underpowered for 12° of wire torsion (twist). In addition, the dual turntable alignment system was not firmly fastened to the sensor, leading to potential forces and moments in other planes of space even though the device was designed to isolate the force and moment in the x-dimension. Only force and torque values in the x-dimension were logged to file; therefore, it would be impossible to determine the overall forces and moments. Movement in the turntable may not isolate the force and torque in the x-dimension; however, it is more realistic clinically than a rigidly held turntable. In a clinical setting the tooth being moved and the adjacent teeth all have a small amount of mobility within the periodontal ligament.

A more significant source of additional forces and moments may be the fact that the bracket was not located directly over the load cell. The distance in the z dimension between the bracket and the origin of the load cell (mm) multiplied by a pre-load force in the y dimension (N) will produce a moment in the x dimension that will add to or subtract from the torque value in the x dimension caused by the twisting of the wire during the experiment. The relative positive or negative effect on torque expression in the x dimension will depend on the direction of the pre-load force. The mobility of the dual turntable alignment system may have dissipated these forces. However, additional forces and torques may have contributed to the bracket debonding that was observed with stainless steel wire at high torsional angles (above 42°). Shear and peel forces may have been created leading to debonding. Other reasons for debonding may have been related to the epoxy resin used for bonding, the degreasing agent and/or the sandblasting process.

5.3 Recommendations for Future Studies

The distance between the bracket and the load cell should be measured in three dimensions. With this information, a transformation matrix could be incorporated into the software to eliminate extraneous forces and torques in order to isolate pure torque in the x dimension caused by wire torsion. Torque expression and bracket deformation could be studied. A range of torsional angles producing 5-20Nmm of torque could be identified for each wire alloy and compared to the results of this study.

There is a lack of evidence regarding the range of clinically effective torque moments.[1] Tooth movement in response to archwire twist and the resultant couple will vary depending on patient factors such as age, sex, bone biology and root form; therefore, clinical trials are required to determine the clinically applicable range of torque moments.[5] There is a need to relate tissue change during tooth movement to the pressure applied to the tooth and supporting structures.[5]

A large, randomized controlled trial is needed in order to determine a safe and effective range of torque moments and the degrees of wire torsion required to attain this range for each alloy. The sample would have to be matched for age, sex, level of bone support, root volume from CBCT measurements and initial inclination. A range of torsional angles thought to be clinically effective for each wire alloy could be applied to patients based on the results of this study (ex. 15-30° for stainless steel wire). Malocclusions of the patients who meet the criteria for the study could be simulated in the Orthodontic Simulator (OSIM) developed at the University of Alberta, in order to approximate the forces and torque moments generated. Root movement could be measured on a CBCT.

5.4 Conclusions

- A systematic review of torque expression revealed that for conventional stainless steel orthodontic brackets with a 0.018" stainless steel bracket slot, the engagement angle ranges from 31° with a 0.016x0.016" stainless steel archwire to 4.6° with a 0.018x0.025" stainless steel archwire. In a 0.022" stainless steel bracket slot, the engagement angle ranges from 18° with a 0.018x0.025" stainless steel archwire to 6° with a 0.021x0.025" stainless steel archwire.
- The results of this thesis reveal that wire alloy is a more important determinant of torque expression than bracket type.
- Torque expression in the Speed bracket was relatively constant beyond 36° of torsion with stainless steel wire.
- Large variability in torque expression within all nine groups of wire and bracket combinations means that torque expression is unpredictable regardless of bracket and wire type. A portion of this variability may be attributed to oversized bracket slots and undersized archwires.
- The range of clinically applicable torque expression spans from 15-30° for stainless steel wire, from 18-42° for TMA and from 21-78° for CuNiTi wire. This extremely large range for CuNiti wire means that torque expression with this alloy will be particularly unpredictable.

5.5 References

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Appendix

Appendix A:

3.2 Materials and Methods

A computerized database search was conducted using Medline, Embase, EBMR (Evidence Based Medicine Reviews), EBM all reviews, PubMed, Scopus and Web of Science up until June 27, 2008.

Terms and their respective truncations used in the literature search (Table) were specific to each database. Searches were made with the help of a senior librarian specialized in Health Sciences. The selection process was carried out together by two researchers.

Table A Database Search and Results

Database	Keywords	Results	Full Articles Retrieved for Evaluation	Articles Selected Based on Final Inclusion Criteria	% of Total Final Selected Articles (11) Found by Database*
Medline	(1) torque OR (2) torque.mp (3) orthodontic brackets (4) orthodontic bracket$.mp. (5) orthodontic wires (6) orthodontic wires$.mp. (1 or 2) AND (3 or 4 or 5 or 6)	176	32	8	73
Pubmed	torque AND (orthodontic brackets or orthodontic bracket* or orthodontic wires or orthodontic wire*)	196	31	8	73
Embase	(1) torque (2) orthod$.mp. 1 AND 2	175	7	3	27
Web of Science	(1) torque* AND orthod*	99	8	4	36
Scopus	(1) torque AND orthod*	359	14	7	70
EBMR	(1) torque.mp. (2) orthod$.mp. 1 AND 2	351	36	9	82
All EBM Reviews	(1) torque.mp. (2) orthod$.mp. 1 AND 2	351	35	9	82

*: Including 2 articles found in hand-search

The following inclusion criterion was chosen to initially select potential articles from the published abstract results of the database search:

- Measurement of torque expression in orthodontic brackets.

Once potentially adequate abstracts were selected, full articles were retrieved for a second selection process. The following additional (final) selection criterion was chosen to select the articles for inclusion in the systematic review:

- *In-vitro* studies measuring torque expression in new stainless steel brackets with a torque-measuring device with straight stainless steel orthodontic wire without second-order mechanics and without loops, coils or auxiliary wires.

Studies measuring friction, bracket deformation/failure, wire torsion, bracket position or tooth morphology without evaluating torque were excluded. Theoretical studies of torque expression without practical application were also excluded.

The QUOROM statement checklist was followed; however, several points did not apply to this systematic review since it was a review of *in-vitro* studies rather than randomized control trials. Validity was assessed by critically examining the torque measuring devices and methodology employed in each study.

Appendix B:

References for Articles in Table 3.1

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Appendix C:

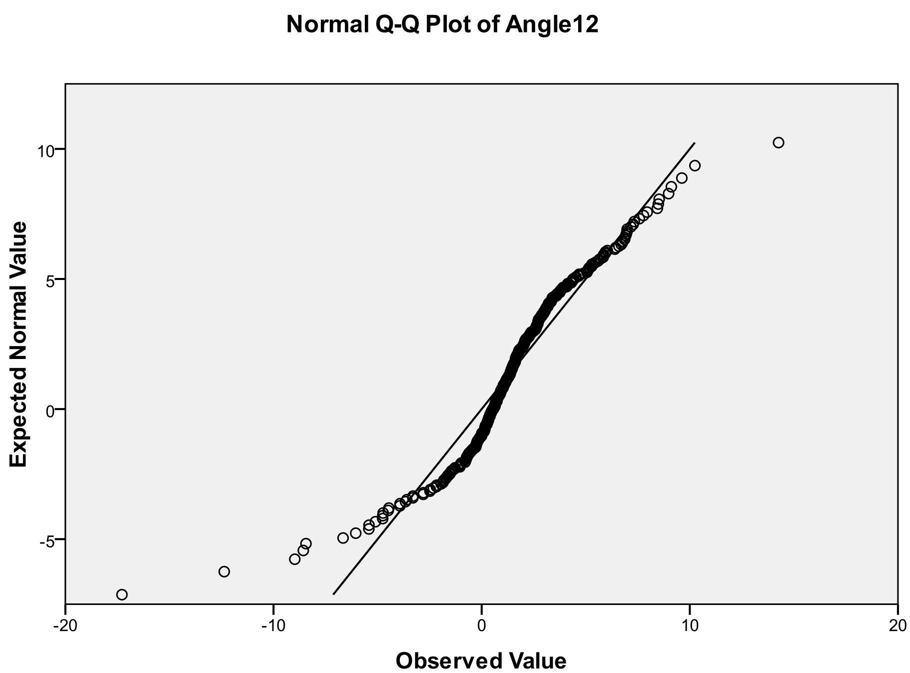

Figure A. Q-Q Plot for 12° of Wire Torsion

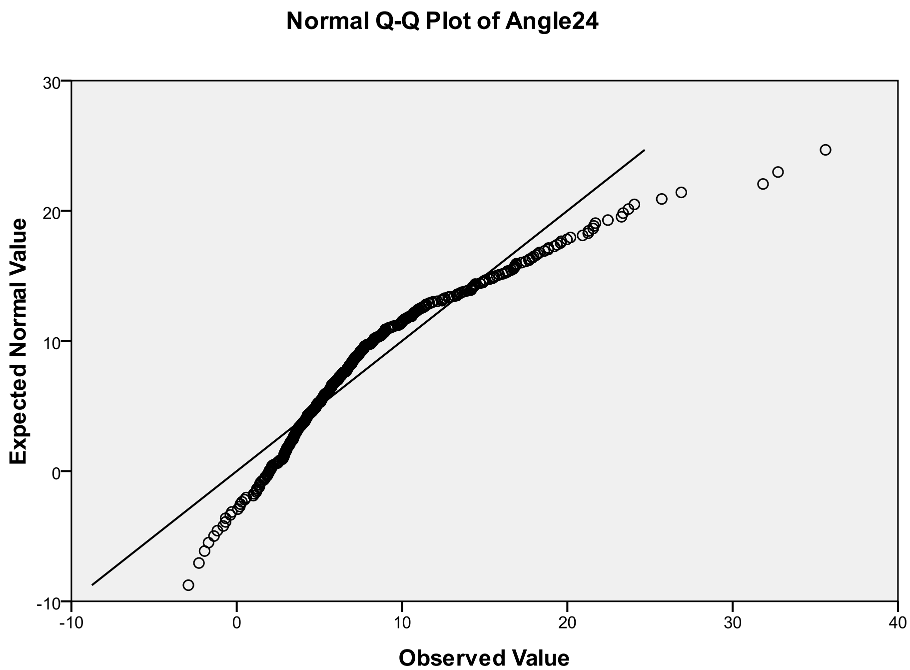

Figure B. Q-Q Plot for 24° of Wire Torsion

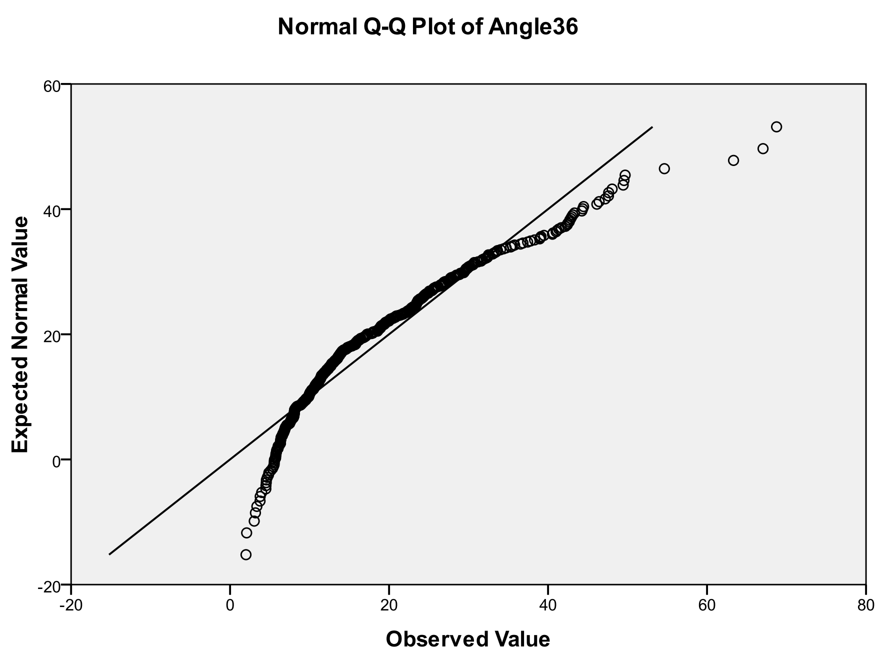

Figure C. Q-Q Plot for 36° of Wire Torsion

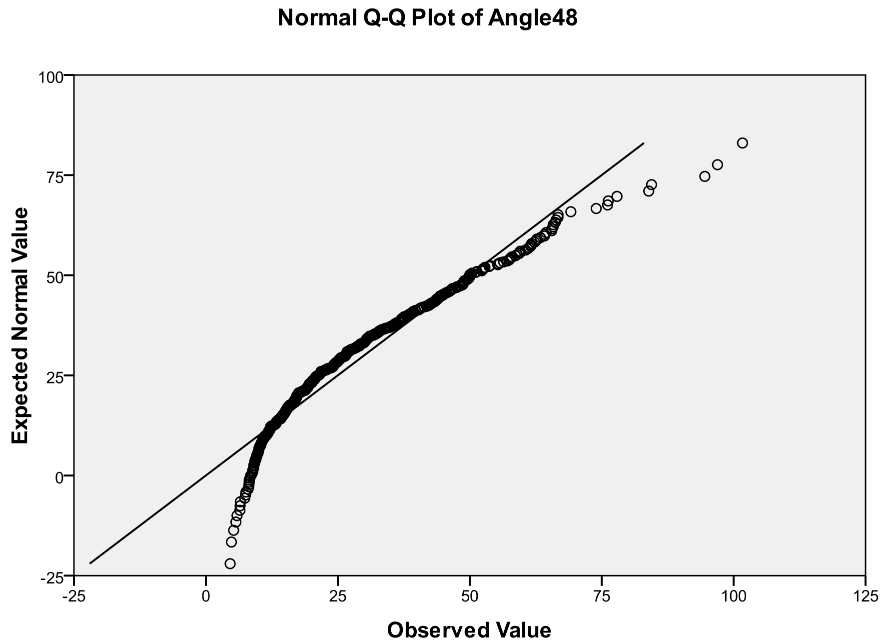

Figure D. Q-Q Plot for 48° of Wire Torsion

Appendix D:

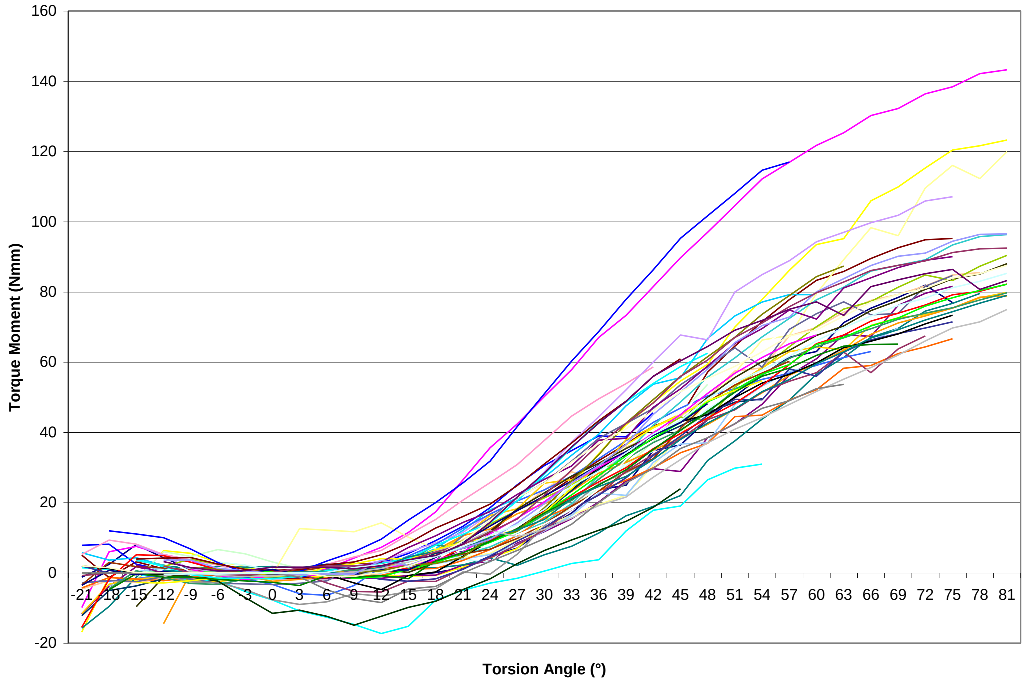

Figure E. Torque Expression for Damon 3MX Brackets with Stainless Steel Wires

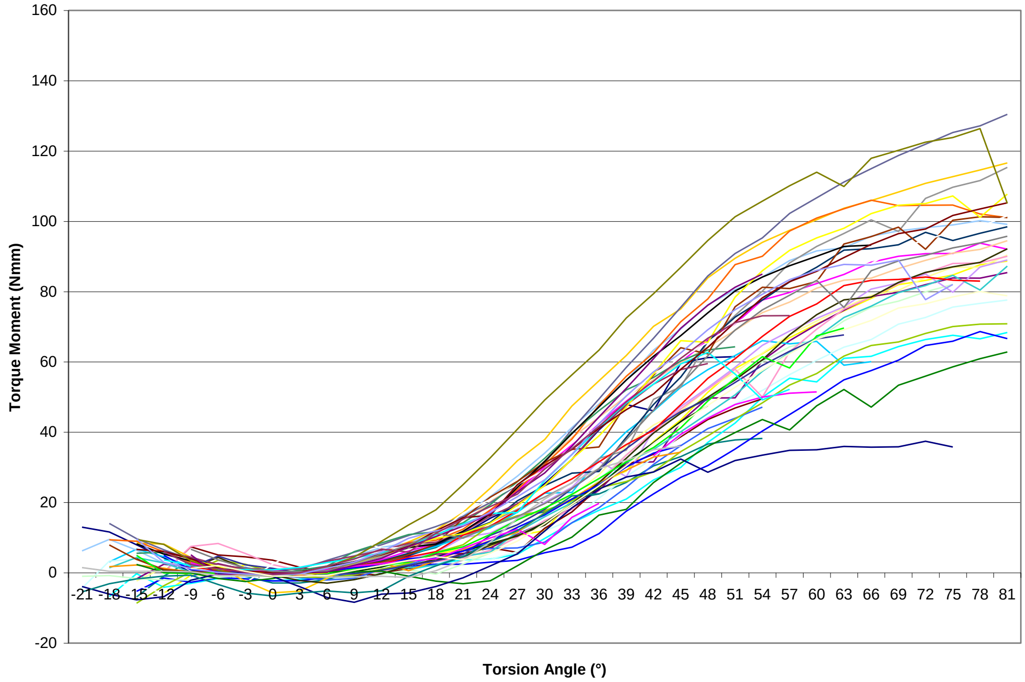

Figure F. Torque Expression for In-Ovation R Brackets with Stainless Steel Wires

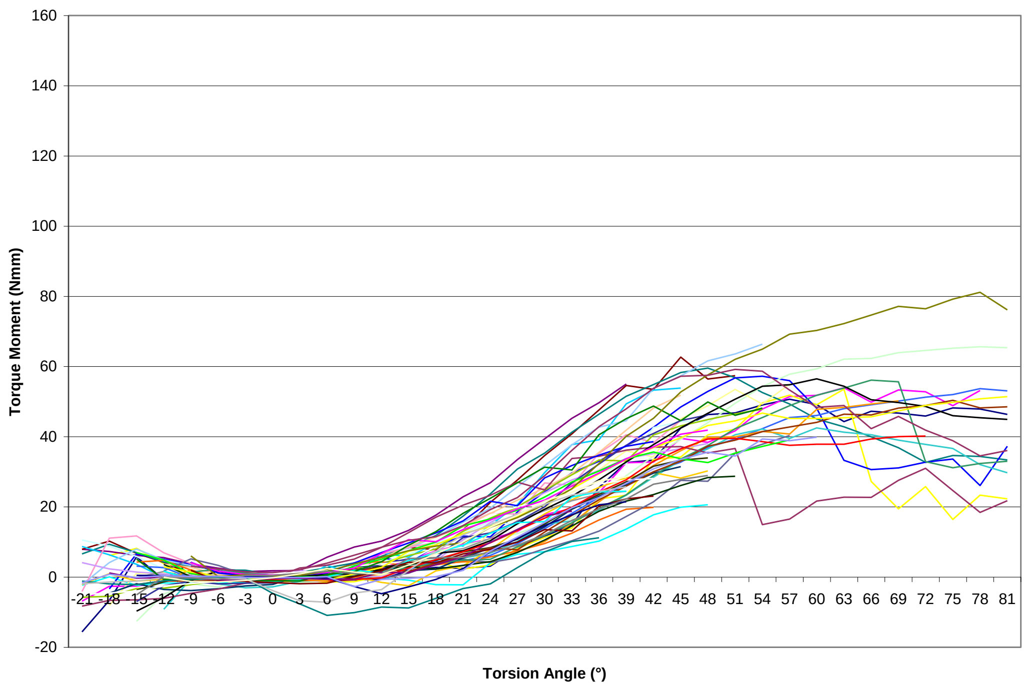

Figure G. Torque Expression for Speed Brackets with Stainless Steel Wires

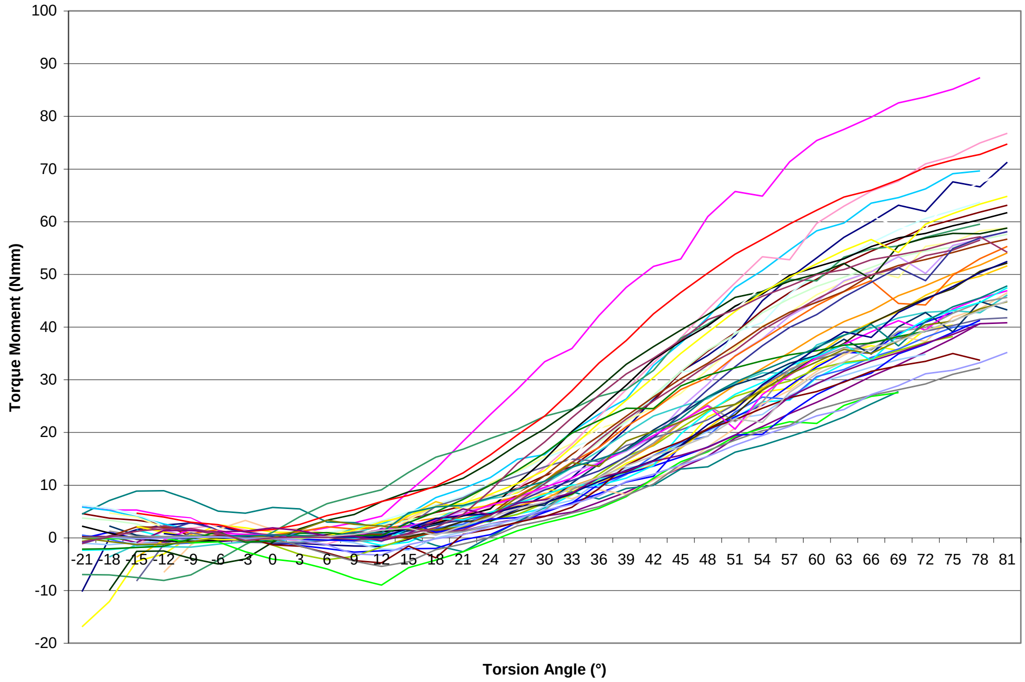

Figure H. Torque Expression for Damon 3MX Brackets with TMA Wires

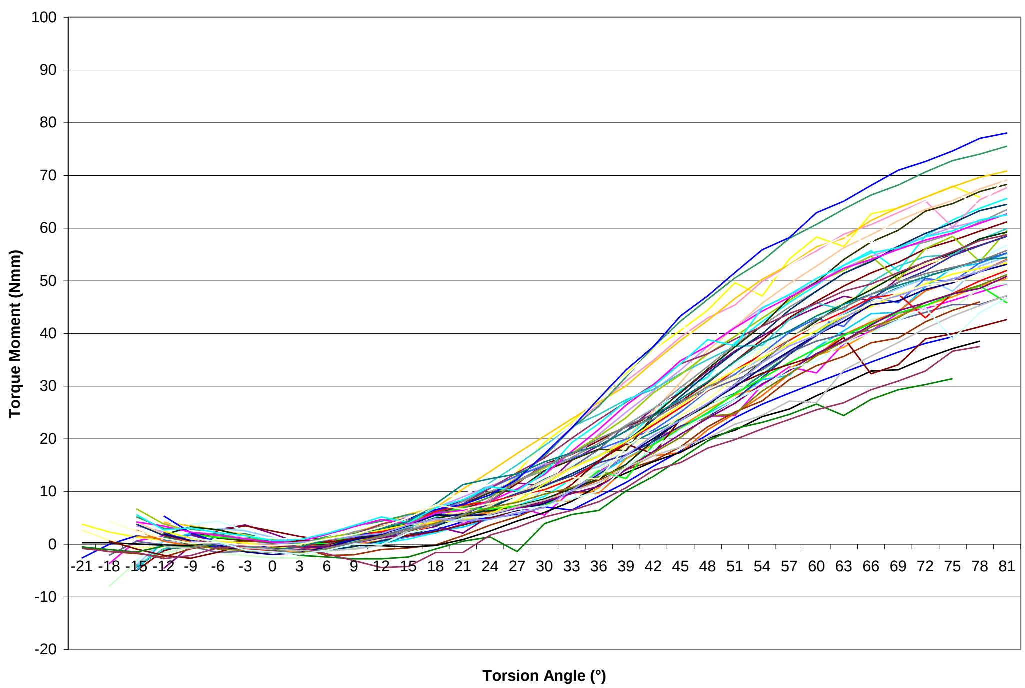

Figure I. Torque Expression for In-Ovation R Brackets with TMA Wires

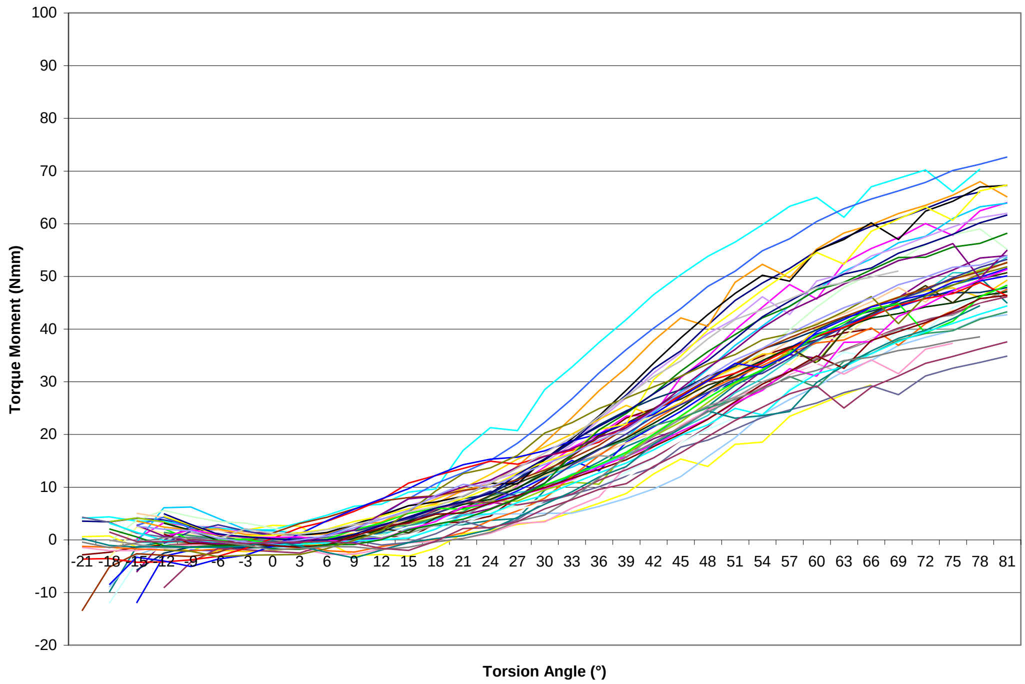

Figure J. **Torque Expression for Speed Brackets with TMA Wires**

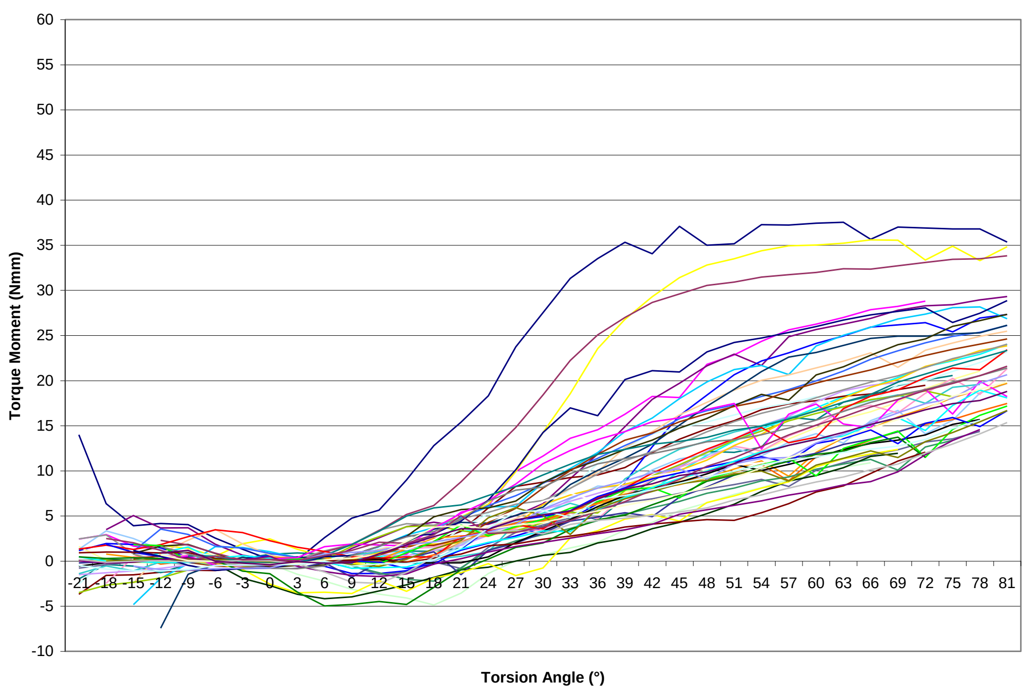

Figure K. **Torque Expression for Damon 3MX Brackets with CuNiTi Wires**

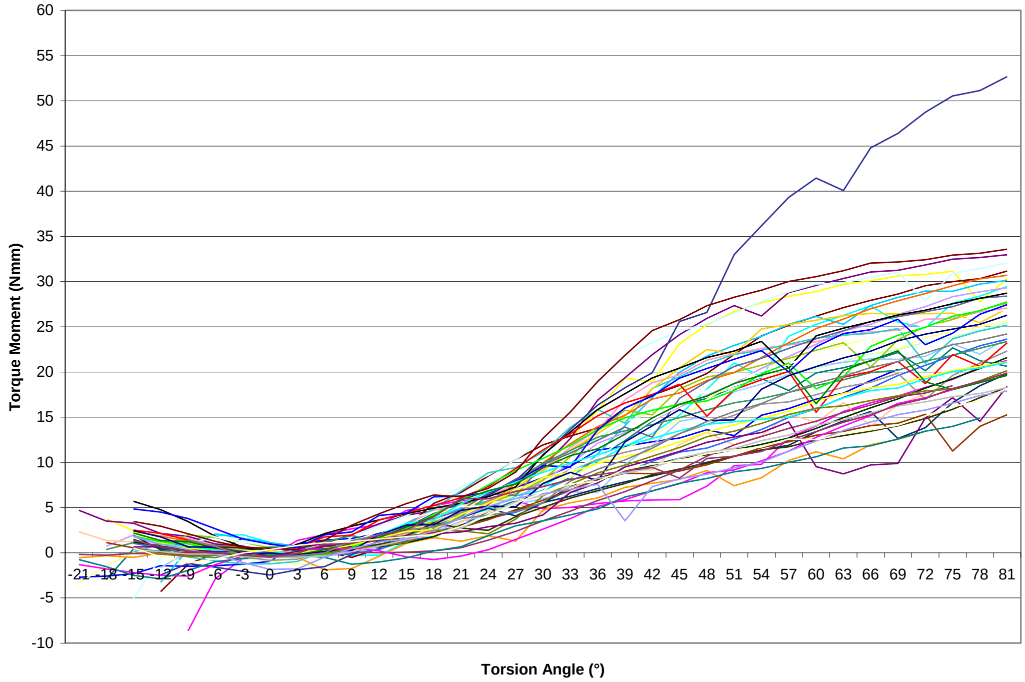

Figure L. Torque Expression for In-Ovation R Brackets with CuNiTi Wires

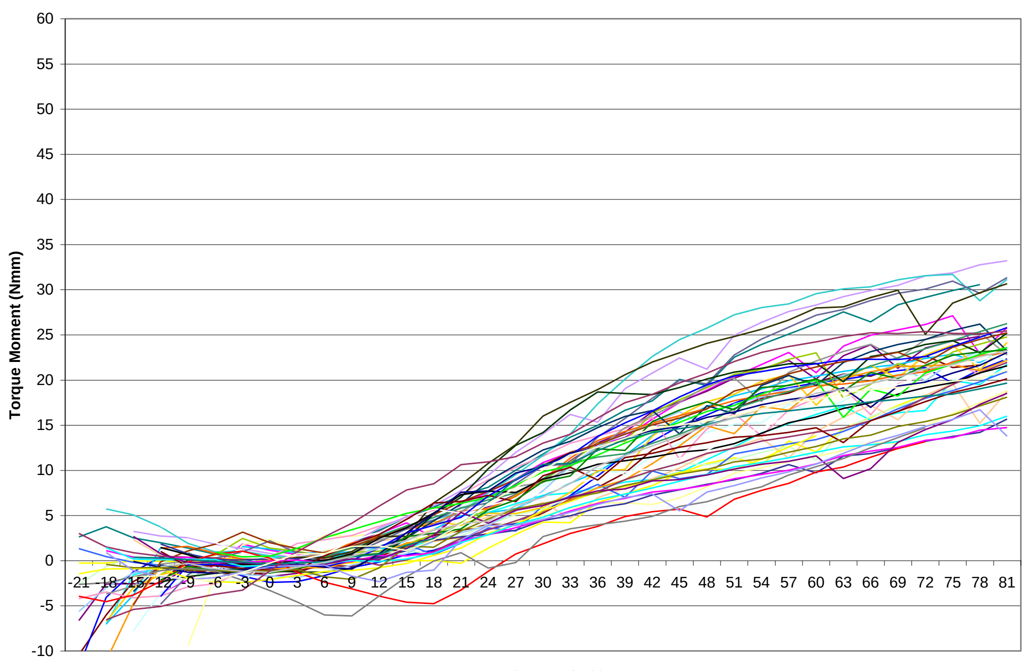

Figure M. Torque Expression for Speed Brackets with CuNiTi Wires