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Torque Expression of Active and Passive Self-Ligating Orthodontic Brackets with Different Stainless Steel Wire Sizes

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Abstract

This investigation aimed to compare engagement angles and torquing moments generated by stainless steel (SS) wires in three active self-ligating (ASL), one passive self-ligating (PSL), and a conventional twin orthodontic bracket system control. Brackets were tested in simulations of buccal and palatal root torque using four sizes of SS wires (16x22, 17x25, 19x25, 20x20 mil). A custom 3D printed testing apparatus was developed to measure torque. After initial engagement within bracket slots, torquing moments increased with progressive rotation of brackets around wires. In general, the PSL and conventionally ligated systems generated significantly larger torquing moments than ASL systems, especially with larger wires and greater degrees of twist. Torquing direction only influenced torque expression with ASL systems. The PSL system demonstrated significantly smaller engagement angles than the ASL or twin bracket systems, especially with larger wires. In addition to ligation modality, other aspects of bracket design likely contribute to these findings.

Keywords

Orthodontic Brackets, Stainless Steel Wire, Third-Order Tooth Movement, Torque, Torque Expression, Ligation, Self-Ligation, Active Self-Ligation, Passive Self-Ligation, Engagement Angle

Summary for Lay Audience

Many people visit the orthodontist for a straighter, more esthetic smile. Achieving desired outcomes requires we properly position the teeth in the mouth in all three dimensions of space. The most common way to accomplish this is using “braces”, which are orthodontic brackets and wires that move teeth into their ideal orientation. One type of tooth movement achieved with braces, known as “torque”, involves changing the inclination of teeth by moving the tooth roots toward or away from the lips or cheeks.

Three basic types of orthodontic brackets are marketed today, differing in terms of how brackets are secured to the wire, or “ligation method”. These different types of ligation method are each suggested to have different advantages, with one particular type (active self-ligation, or ASL) suggested to add torque to teeth better than other methods. To test this claim, five different bracket systems representing all three ligation methods were tested by twisting brackets from 0 to 45 degrees around a section of orthodontic wire and measuring resulting torquing moments. This was repeated in both directions. Four common sizes of stainless steel orthodontic wires were considered. Resulting moments were compared between different bracket types, wire sizes, directions of twist, and to existing literature to determine if one ligation method was superior in producing torque.

In general, with greater degrees of twist, torquing moments increased for all bracket systems tested, once the wire and bracket engaged one another. Likewise, for a given degree of twist, torquing moments tended to be higher for larger wires than smaller wires. However, comparing different ligation types, tested ASL groups seemed to generate lower moments than other tested brackets, despite purported benefits of this ligation type. Direction of rotation affected only ASL systems and not other ligation method groups. These findings are likely the result of not only ligation modality, but other aspects of bracket design as well. A prudent clinician should possess a full understanding of the appliances employed in clinical practice to treat patients efficiently. We hope orthodontists can use this information to optimize use of bracket systems and improve outcomes of orthodontic treatment.

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List of Abbreviations

µm	Micrometer
3D	Three-dimensional
A-Emp(B)	Empower®2 Metal Interactive brackets simulating Buccal Root Torque
A-Emp(P)	Empower®2 Metal Interactive brackets simulating Palatal Root Torque
ASL	Active Self-Ligation
A-Spd(B)	Speed System™ brackets simulating Buccal Root Torque
A-Spd(P)	Speed System™ brackets simulating Palatal Root Torque
A-Vic(B)	3M™ Victory Series™ Active SL brackets simulating Buccal Root Torque
A-Vic(P)	3M™ Victory Series™ Active SL brackets simulating Palatal Root Torque
cNmm	CentiNewton Millimeters
C-Vic(B)	3M™ Victory Series™ Twin brackets simulating Buccal Root Torque
C-Vic(P)	3M™ Victory Series™ Twin brackets simulating Palatal Root Torque
g	Grams
GUI	Graphical User Interface
kN	Kilonewton
MBT	Mclaughlin, Bennett, Trevisi prescription
Mc	Moment of a Couple
Mf	Moment of a Force
mil	Thousandths of an inch

MIM	Metal Injection Molding
mm	Millimeter
N	Newtons
NiTi	Nickel Titanium
Nmm	Newton Millimeters
OMSS	Orthodontic measurement and simulation system
P-Dmn(B)	Damon™ Q2 brackets simulating Buccal Root Torque
P-Dmn(P)	Damon™ Q2 brackets simulating Palatal Root Torque
PSL	Passive Self-Ligation
SD	Standard Deviation
SL	Self-Ligation
SS	Stainless Steel
TMA	Titanium Molybdenum Alloy

Chapter 1

1 Literature Review

1.1 Introduction

Orthodontic treatment is commonly sought for children, teens, and adults alike in contemporary society. According to Proffit et al[1], the primary goal of orthodontic treatment today is normal soft tissue proportions and adaptations, with secondary goals being a functional occlusion. To achieve both esthetic and functional goals of treatment, the orthodontist must pay careful attention to the three-dimensional orientations of the patient's teeth, including intra-arch and inter-arch dental relationships. More specifically, according to the American Board of Orthodontics[2], orthodontic treatment should aim to achieve eight main criteria centered on dental and occlusal relationships: ideal tooth alignment, ideal marginal ridge alignment, correct buccolingual inclination, adequate posterior occlusal contacts, correct anteroposterior occlusal relationships, ideal overjet, tight interproximal contacts, and correct root angulation. Obtaining adequate buccolingual inclination, also known simply as inclination, or alternatively as torque, is a frequent challenge. Achieving the goals of orthodontic therapy, including correct torque, requires the orthodontist have a full understanding of the mechanism by which orthodontic appliances produce orthodontic tooth movement and the limitations of the various appliances available.

1.2 Biomechanical Principles of Tooth Movement

Orthodontic tooth movement is dependent upon the application of an orthodontic force to a tooth, thereby initiating stresses and strains in the periodontium and in turn, a cascade of biological events that lead to orthodontic tooth movement. Understanding basic mechanical principles is imperative to understand how orthodontic appliances produce orthodontic tooth movement in this manner, and thus a brief examination of fundamental mechanical concepts is the first subject of this review.

The first concept to be introduced is that of a force, or an action applied to a body.[3] Specifically, a force is a load applied to an object that tends to move the object to a different position in space.[1] More intuitively, a force can be described as a push or pull that tends to change an object's velocity. Critically, these can be represented as vectors, indicating forces have both magnitude and direction.[3,4] In turn, the point of application of a force indicates the point of contact between a body being moved by the force and the applied force, and the line of action indicates the line along which the force acts.[4] These features of a force may be represented by diagrams, as in Figure 1.

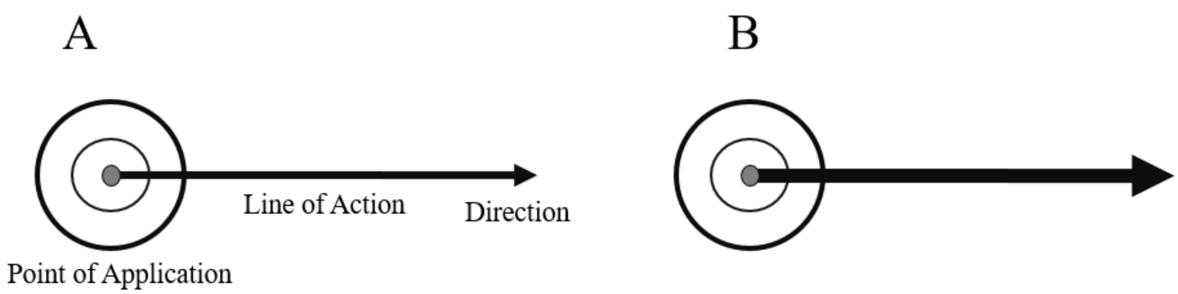


Figure 1: Forces are vectors characterized by a point of application, line of action, and direction. Magnitude may be indicated by thickness of the force vector, with (A) demonstrating a smaller force and (B) a larger force

The way in which an object moves in response to an applied force depends on the relationship between the line of action of the applied force and, if the object is a free body in space, the object's center of gravity, which describes a point in which all of the object's weight is concentrated.[4,5] If a force is applied with a line of action through a free body's center of gravity, the object will tend to move linearly, or translate. In other words, all points of the body will move in the same direction by the same amount.[5]

A tooth, however, does not represent a free body in space - teeth are restrained by the periodontal apparatus. As such, a center of gravity does not apply to the dentition. Instead, a center of resistance can be described for teeth, where the center of resistance is analogous to the center of gravity, but for a restrained body. [1,3–5] Importantly, the position of a tooth's center of resistance is located along the root of the tooth (Figure 2), with the precise location varying with several factors, including root length, number of roots, and alveolar bone height.[3–5] A force with a line of action through a tooth's center of resistance will tend to translate the tooth.[3–5]

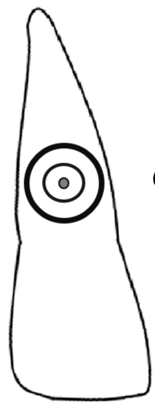


Figure 2: The Center of Resistance of a tooth is located along the root of the tooth, with the exact position varying with numerous factors. A force with a line of action applied to this point will tend to produce translation

If the line of action of an applied force does not pass through the center of resistance of an object, the force will tend to produce rotation of the object in addition to translation[3–6], where rotation involves movement of the body whereby no two points of the body move in the same direction by the same amount.[5] The object will appear to rotate around its center of resistance while the center of resistance simultaneously translates.[1,5] Put another way, the rotation produced will occur around a point termed the center of rotation, which can be defined as a point about which the object appears to have rotated, as determined by the initial and final position of the object.[4]

The tendency of a force to produce rotation of an object is termed the moment of the force (Mf), and is equal to the magnitude of the force multiplied by the perpendicular distance between the line of action of the force and the center of resistance of the object[1,3–5], as shown in Figure 3. In orthodontics, moments are often described in terms of Newton millimetres (Nmm). The magnitude of the tendency to rotate is increased equally by either increasing the force applied, or by applying the force so the line of action is further from the centre of resistance.

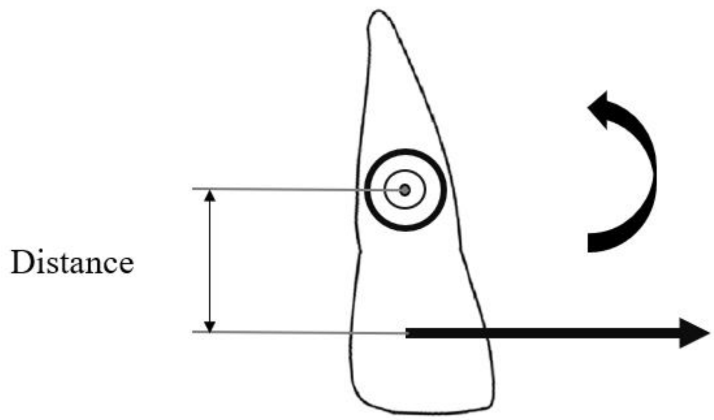


Figure 3: A force applied a distance from the Center of Resistance will tend to produce rotation. The tendency to produce rotation is known as Moment of the Force, which is equal to the magnitude of the force multiplied by the distance between the Line of Action of the force and the Center of Resistance of the object

In turn, if two forces are applied to an object simultaneously, more complex motion can be produced. For instance, consider a couple. A couple consists of two non-collinear forces equal in magnitude but opposite in direction.[3–5] The moment of a couple (Mc), or the tendency of a couple to produce rotation, is equal to either the sum of the two moments of the forces, or to the magnitude of one of the forces multiplied by the perpendicular distance between the two forces[3–5], as shown in Figure 4. A couple will produce rotation of a body around its center of resistance, with no translation.[3–6] In other words, a couple applied to an object will tend to produce pure rotation of the object with a center of rotation coincident with the center of resistance of the object. Critically, where a couple is applied to an object is irrelevant – the net effect of rotation about the center of resistance is the same, no matter where on the object the couple is applied.[3–5]

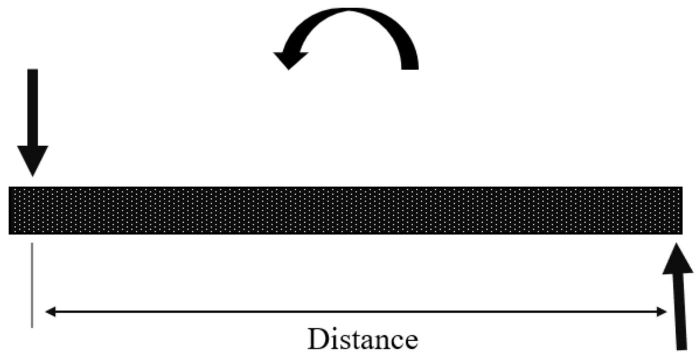


Figure 4: A couple involves two non-collinear forces equal in magnitude but opposite in direction. The tendency of a couple to produce rotation is the Moment of a Couple and is equal to the magnitude of one of the forces multiplied by the perpendicular distance between the two forces

Orthodontic forces are classically applied to the crowns of teeth while their center of resistance is located along the root, therefore moments are frequently produced. By applying mechanical concepts and implementing combinations of moments and forces to teeth using orthodontic appliances, the orthodontist can control the center of rotation of a tooth, thereby controlling the type of tooth movement achieved.[4,6] Indeed, applying various combinations of both a force and a couple to a tooth facilitates producing any type of tooth movement, with any center of rotation.[1,4]

1.3 Categories of Tooth Movement

The variety of tooth movements possible with an orthodontic force system that employs a combination of a force and couple can be broken down into five basic categories of tooth movement characterized by the position of the center of rotation: translation, uncontrolled tipping, controlled tipping, root movement, and pure rotation.[1,3,4,6] Roughly, if the center of rotation produced is at an infinite distance away from the center of resistance with the root apex and crown moving in the same direction by the same amount, translation is occurring (Figure 5A), whereas if the center of rotation is slightly apical to the center of resistance, a tooth can be described as exhibiting uncontrolled tipping (Figure 5B).[3,4] If the center of rotation is at the apex of the tooth, controlled tipping (Figure 5C), also known as crown movement, is being produced, while a center of rotation at the incisal or occlusal aspect of the tooth is associated with root movement (Figure 5D).[1,3,4,6] Finally, if the center of rotation is coincident with the center of resistance, pure rotation or pure tipping has occurred (Figure 5E).[3,6]

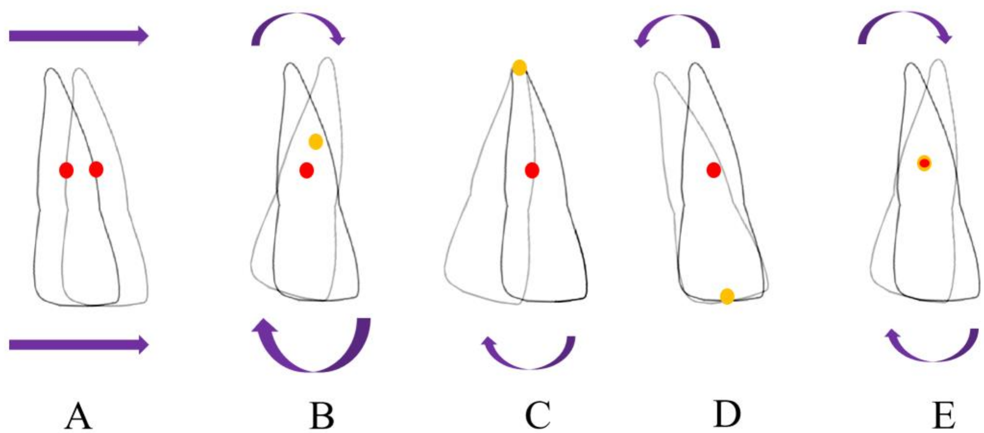


Figure 5: Categories of orthodontic tooth movement, including translation (A), uncontrolled tipping (B), controlled tipping (C), root movement (D), and pure rotation (E). Red dots represent the center of resistance while orange dots represent the center of rotation

Importantly, the location of the center of rotation, and subsequently whether translation, root movement, controlled tipping, uncontrolled tipping, or pure rotation occurs is determined by the ratio between the magnitude of the couple and the magnitude of the force applied at the bracket, or in other words, the moment-to-force ratio.[4,6] Alternatively, the ratio of the Mc to the Mf can be used to represent how a tooth will move in response to an applied force and a couple.[1] Overall, when only a force is applied at a bracket with no couple ($Mc/Mf = 0$), uncontrolled tipping is produced with center of rotation just apical to the center of resistance.[1,3,4] If the moment of a couple is present but less than the moment of the force ($0<Mc/Mf<1$), controlled tipping or crown movement occurs[1,3], and if the moment of the couple and moment of force are balanced (Mc/Mf=1), translation occurs.[1,3,6] Finally, if the moment of the couple is greater than the moment of the force (Mc/Mf >1), root movement results[1,3], whereas if only a couple is applied to the tooth, pure rotation occurs.[3,6]

The different categories of tooth movement can also be described in orthodontics in terms of first, second, and third order movements, depending on the orientation of the movement in three-dimensional space. Labio-lingual movements of a tooth from the occlusal view are described as first order movements (Figure 6A), mesiodistal root paralleling and occlusogingival levelling as viewed from the buccal or lingual view are described as second order movements (Figure 6B), and finally, buccolingual and labiolingual root positioning as viewed from the mesial or distal view are described as third order movements (Figure 6C).[3] In turn, the term “angulation” or “tip” is often used to describe the mesiodistal orientation of the tooth, while “inclination” or “torque” is used to describe the labiolingual or buccolingual orientation.[7] Each order of movement is critical in achieving an esthetic and functional orthodontic result, thus the orthodontist must be proficient in achieving precise control over each aspect of orthodontic tooth movement.[1,7,8]

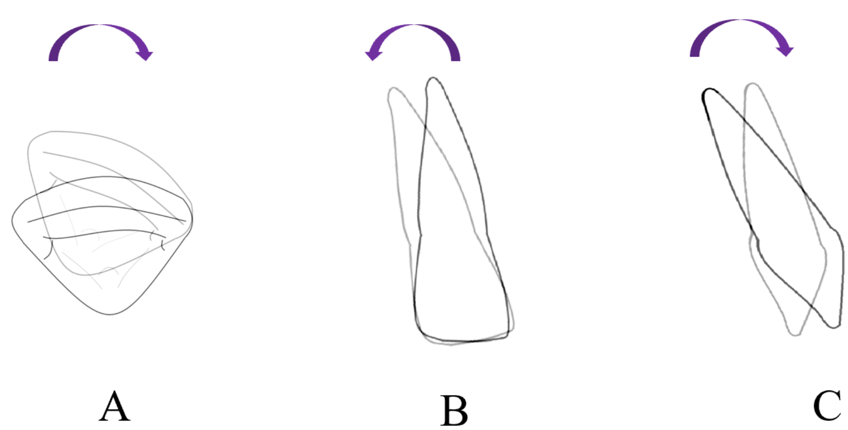


Figure 6: Categories of orthodontic tooth movement, including first order movements (A), second order movements (B), and third order movements (C)

1.4 Contemporary Orthodontic Appliances

Several strategies and appliances are available to achieve detailed control of the various orthodontic tooth movements during orthodontic treatment, but fixed appliances using brackets bonded to the teeth along with orthodontic archwires are a classic approach.

Specifically, contemporary fixed orthodontic appliances are characterized by brackets bonded to the teeth containing a slot into which orthodontic archwires are inserted. A variety of bracket systems are available on the market today, differing in terms of bracket size and shape, slot dimension, bracket material, and ligation method, among others, with different systems selected depending on clinical need, as well as operator and patient preference. For instance, modern appliances include both conventional brackets into which archwires are held using stainless steel or elastomeric ligatures, and self-ligating (SL) brackets that employ clips and springs to hold the archwire in the slot, each in turn available in multiple different materials, including metal and ceramic options (Figure 7).

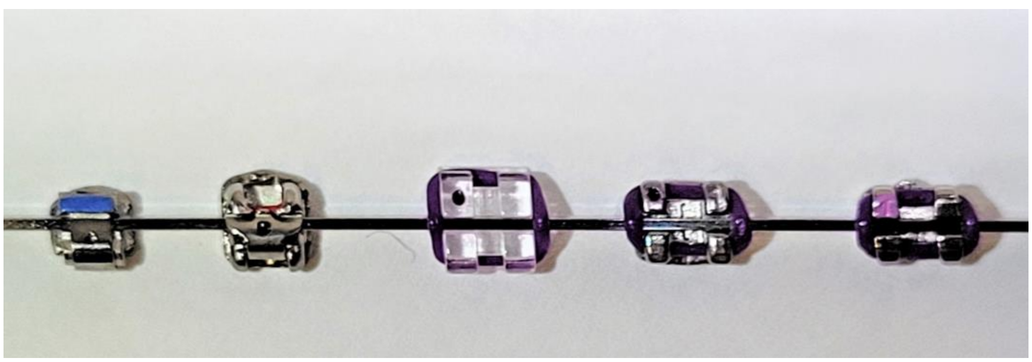


Figure 7: A selection of contemporary orthodontic brackets, including conventional and self-ligating brackets, as well as metal and ceramic brackets, mounted on a stainless-steel orthodontic wire

Moreover, contemporary fixed orthodontic appliances today are available in two slot heights, specified in inches: 0.018 or 0.022 inches. Note that alternatively, orthodontic dimensions are commonly reported in mils, or thousandths of an inch, where 0.018 inches is equivalent to 18 mils, or millimetres, where 0.018 inch is equivalent to 0.4572 mm. Mils are perhaps the most common unit used in regard to orthodontic appliances thus will be the unit employed throughout this document.

In turn, first, second, and third order movements are pre-programmed into the modern-day fixed appliance, hence appliances may be referred to as pre-adjusted appliances, with various prescriptions available depending on patient needs and operator preferences. Likewise, various archwire materials and dimensions are utilized in clinical practice today.

Importantly however, regardless of specific fixed appliance employed, the basic function of the various appliances is similar: engagement of an orthodontic archwire in the bracket slot enables application of two point contacts to teeth and thus application of both a force and a couple to the tooth, and thereby precise control of both the center of rotation and tooth movement.[1] For example, to achieve third order (torque) movements of the teeth, a square or rectangular wire is twisted within the bracket slot until it engages the slot. When the wire contacts the walls of the slot, it undergoes torsion to generate a couple to produce root movement (Figure 8). It follows that torque is typically quantified by measuring the angle of the degree of twist in the orthodontic wire relative to the bracket slot.[3]

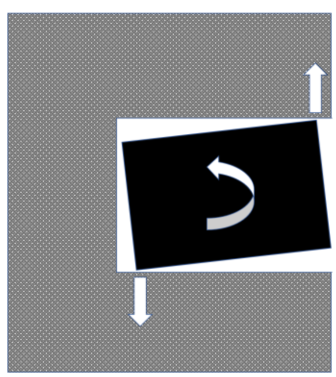


Figure 8: Pictorial representation of a cross-section of wire twisting within a bracket slot to produce torque, as viewed from a mesial or distal view. The grey patterned area represents the bracket, black represents a wire engaged in the slot, central white arrow represents direction of torsion of the wire, and the vertical white arrows represent contact points of the wire against the slot walls, producing a couple that would generate torque

1.5 Stages of Comprehensive Orthodontic Treatment

Treatment using a pre-adjusted appliance characteristically progresses from small round wires in initial phases of treatment through larger rectangular wires in later stages of care. Wire bends and auxiliary appliances may be employed throughout treatment to assist the appliance in meeting treatment goals.

Specifically, as reviewed in Proffit et al[1] and McLaughlin et al[8], in the first phase of comprehensive orthodontic treatment with contemporary fixed appliances, known as alignment and levelling, the primary goals of care are to bring the teeth into alignment and correct vertical discrepancies by levelling the arches. In the alignment stage, light continuous forces are provided by small round wires with at least 2 mil of clearance between wire and bracket slot, with tipping as opposed to root movement a main focus. Indeed, root movement is avoided in the initial phases of treatment in many cases. Small round nickel titanium (NiTi) wires are typical in this initial phase. For levelling, vertical corrections are generally achieved with a stiffer wire than used for aligning. Round stainless steel (SS) archwires are a typical selection. When alignment and levelling is complete, the teeth should be aligned such that a rectangular wire can be placed without excessive force.

The second phase of comprehensive orthodontic treatment aims to correct molar relationships and close spaces if present. This correction may involve growth guidance, or differential anteroposterior movements of the maxillary and mandibular dentition. Space closure mechanics may be employed as needed, and the use of auxiliaries, such as inter-arch elastics, is common. Typical archwires used during the second phase of comprehensive treatment are rectangular SS wires or rectangular titanium molybdenum alloy (TMA) wires, with incorporated elastics, springs, and closing loops as needed, depending on the clinical situation. As treatment progresses through these larger stiffer rectangular wires, the wire engages in the bracket slot, expressing the pre-programmed features of the appliance, including torque. After rectangular wires have been in place for an extended period, the pre-programmed prescription of the appliance is largely expressed. Upon completion of the second phase of treatment, the teeth should be well aligned, extraction spaces closed, roots should be reasonably parallel, and a class I buccal occlusion should have been achieved, although some refinements will characteristically be needed in the subsequent final stage of treatment. For instance, correction of side-effects of anteroposterior correction and space closure, such as lingual tipping of incisors, may still be required at the completion of the second phase of treatment.

Finally, the third and final stage of comprehensive orthodontic treatment, known as finishing, aims to level marginal ridges, obtain precise tooth positioning, and overcome any other discrepancies. An emphasis is placed on achieving correct third order tooth positions in this phase of care, although with modern pre-adjusted appliances, only moderate additional torque should be needed. The archwires used to accomplish finishing should fill the bracket slot but have appropriate flexibility to accomplish the desired movements. Typical choices may include a combination of large rectangular NiTi, TMA, and SS wires, depending on clinical needs. Regarding torque control during this phase of care, for an 18 mil appliance, rectangular SS wires offer excellent properties for expression of torque, whereas in a 22 mil appliance, large dimension rectangular braided SS or TMA wires are typical choices. Auxiliary appliances, such as a piggyback arch, may be needed to idealize inclination, and torquing bends may be employed as needed.

1.6 Engagement Angle

A critical factor in producing third order tooth movements with a pre-adjusted appliance in any phase of care is the engagement angle of the wire within the slot. Specifically, when an orthodontic wire is inserted into a bracket slot, if the wire is undersized relative to the slot, it will be able to rotate within the slot before it contacts the walls of the slot. Only when the wire contacts the walls of the slot will it begin to undergo torsion and generate a couple to produce root movement. The angle of freedom of the wire in the slot is known as the “engagement angle”, or “slop” (Figure 9). In turn, the total amount of freedom of wire within the slot is often referred to as “torsional play”. Critically, engagement angle and play is influenced by bracket slot size and various features of the orthodontic archwire employed, including edge bevel of the archwire and wire dimension, among many other factors. Altogether, increased play is associated with reduced torque expression with an orthodontic appliance.

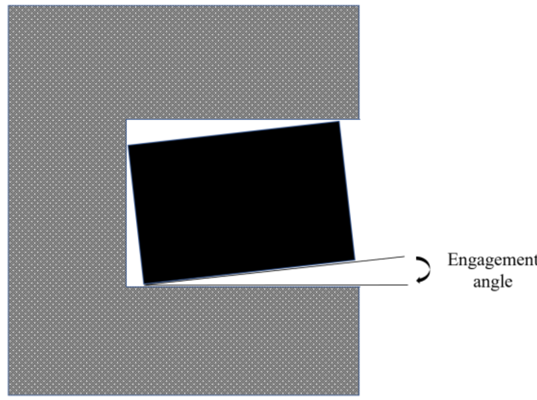


Figure 9: Pictorial representation of a cross-section of wire twisting within a bracket slot, as viewed from a mesial or distal view of the bracket. The grey patterned area represents the bracket, black represents a wire engaged in the slot. Engagement angle is outlined within the image

Interestingly, engagement angles and torsional play in an appliance can be determined indirectly by various methods, such as via a calculation taking into account edge bevel of the archwire, wire dimensions, and bracket slot height[9,10], or play can be measured directly.[11] Measuring torsion in the archwire when a torquing moment is first observed, as in Badawi et al[12], is a particularly reliable method given this technique utilizes real data points to determine the engagement angle.

Theoretical engagement angle with a typical combination of archwire measuring (in terms of height x depth) 19 x 25 mils in a 22 mil slot is generally estimated at about 7-9 degrees, whereas the engagement angle of a 17 x 25 mil wire in an 18 mil slot is estimated at about half that amount.[1,13] However, clinical measurements of engagement angles and torsional play often find more freedom between wire and bracket slot than is expected based on theory[11,12,22–24,14–21] and thus reduced torque control than anticipated in often observed. For instance, Dalstra et al[24] investigated the degree of twist needed to generate a moment in both an 18 and 22 mil slot using a 17 x 22 mil and 19 x 25 mil SS wire respectively and found between 19.8 and 36 degrees of play depending on the bracket system under investigation, which was larger than anticipated according to theoretical calculations.

These observations are hypothesized to result from numerous sources, including undersized archwires versus manufacturer specifications, oversized slots versus manufacturer specifications, wire edge beveling, among other factors.[11,12,22–25,14–21] Contrarily, some investigations have noted play is less than theory suggests for some bracket systems.[12]

1.7 Torque Values in Orthodontics

1.7.1 Threshold Torquing Moment Values

Ultimately, to produce any type of orthodontic tooth movement in any stage of treatment, a threshold magnitude of forces and moments must be applied to the tooth. This is partially due to stabilization effects of the periodontal ligament that must be overcome by orthodontic forces.[1] Regarding torque, clinically effective torque has been suggested to range between 5 and 20Nmm.[20,26,27]

1.7.2 Risks of Torque Application

It is important to note that application of moments and forces does not come without risk, even when forces and moments applied are within the physiological range, although in general, the greater the forces and moments applied, the more risk of adverse outcomes. For example, although the cause of root resorption during orthodontic treatment is not completely understood and is likely multifactorial, torquing movements are a known risk factor, with higher magnitudes of torque associated with increased resorption. For instance, Casa et al[28] applied lingual root torque moments of 300 and 600 centiNewton millimeters (cNmm) to upper first premolars and compared resorption patterns to control teeth that were not subject to torque and noted teeth that received higher magnitudes of torquing moments demonstrated greater root resorption, both in terms of width and depth, with resorption concentrated in the apical third of the root. Likewise, more recently, Bartley et al[29] compared the degree of root resorption after 2.5 and 15 degrees of buccal root torque over a four week period and noted root resorption was evident in both groups and concentrated in areas of compression, but significantly more resorption was noted in the apical regions of the groups receiving higher magnitudes of torque, although the authors concluded mechanical factors alone were inadequate to explain the etiology of resorption.

Evidently, care must be taken to control torquing moments when providing orthodontic treatment such that a minimum threshold is reached without utilizing excessive forces, thus the prudent orthodontist must have a full understanding of factors that influence torque expression with the appliances they employ so as to maximize efficiency while minimizing risk of harm.

1.8 Factors Influencing Torque Expression

Various factors influence how effectively pre-adjusted appliances produce forces and couples and consequently tooth movement. For instance, the mesiodistal width of the bracket slot is key to produce first-order movements, while mesiodistal width of the bracket as well as the vertical dimension of the bracket slot are key in producing second-order tooth movements.[3] Properties of the orthodontic archwire, bracket positioning, and tooth morphology among others, will also have an influence on the function of the pre-adjusted appliance.[3] Considering torque, according to Proffit et al[1], three main categories of factors determine torque expression in a pre-programmed rectangular slot: (1) tightness of fit between the archwire and the bracket slot (engagement angle), which in turn is affected by numerous variables; (2) the properties of the wire, especially torsional stiffness; and (3) inclination of the bracket slot relative to the wire, which can be influenced by factors such as tooth anatomy and bracket torque prescription. More specifically, factors that may influence torque expression using pre-adjusted orthodontic appliances are reviewed below.

1.8.1 Bracket Slot Dimensions

Bracket slot size, especially slot height, may have an effect on torque control with orthodontic appliances. As noted, contemporary pre-adjusted appliances are available in two slot heights: 18 and 22 mil. More precisely, the typical dimensions of an 18 mil appliance in terms of height x depth is 18 x 25 mil (0.018 x 0.025 inch or 0.46 x 0.64 mm), while a 22 mil appliance generally features a slot 22 x 28 mil (0.022 x 0.028 inch or 0.56 x 0.71mm).

Clinically, 22 mil appliances have been found by some authors to underperform in terms of torque expression as compared to 18 mil appliances when considering the American Board of Orthodontist's Objective Grading Scheme.[30] When considering large dimension stainless steel (SS) wires, the rationale for the observation 22 mil appliances may underperform is that large SS wires required to fully engage the 22 mil slot appliance are so limited in springiness and range in torsion that effective torque control is impossible.[1] To overcome these limitations, the strategy of utilizing an undersized SS wire may be adopted. Specifically, in the 22 mil appliance, undersized SS wires of 19 x 25 mil (3 mil of clearance) are often the largest SS wire employed, whereas with an 18 mil appliance, full sized SS wires as large at 17 x 25 mil (1 mil of clearance) are often utilized.[1] However, despite this modification, torque efficiency is often found to be higher in an 18 mil appliance as compared to a 22 mil appliance. For instance, when comparing typical wires of 17 x 25 mil SS in an 18 mil bracket to 19 x 25 mil SS wires in an 22 mil bracket, Sifakakis et al[31] found the maximum moments generated with 15 degrees of twist in the 18 mil appliance were statistically significantly higher than those generated with the 22 mil appliance with the same degree of twist. Similar findings were reported in subsequent studies comparing SS wires in 18 and 22 mil appliances.[32,33] The rationale for these observations is that there is less play in 18 mil appliances with full sized SS wires as compared to 22 mil appliances with undersized SS wires.[31,32] The clinical significance of increased play and reduced moments in a 22 mil slot appliance with undersized SS wires can be questioned however, given the finding 18 and 22 mil appliances using 16 x 22 mil and 19 x 25 mil SS wires respectively as a final stage of treatment produce treatment results that are not statistically or clinically significantly different.[34,35]

Alternatively, to overcome the weaknesses of a full sized or undersized SS wire in a 22 mil slot, full sized wires of alternative materials may be used aside from SS, such as titanium molybdenum alloy (TMA) or nickel titanium (NiTi). Indeed, full sized (21 x 25 mil) TMA and NiTi wires are often used in later stages of treatment with a 22 mil slot appliance.[1] Critically, if full sized wires are employed (17 x 25 mil in the 18 mil appliance and 21 x 25 mil in the 22 mil appliance), a systematic review conducted by Archambault et al[20] found that there was no statistical difference in engagement angle between 18 mil and 22 mil appliances, contributing to improved torque control with a 22 slot appliance.

It follows that the dimensional accuracy of fabrication of the slot can also influence torque expression. The dimensions of a given bracket slot can be determined through direct measurement of the slot[36–38], or bracket slot height can be estimated using a formula that describes a relationship between bracket slot height, wire dimension, wire edge bevel, and torsional play as in Meling et al.[10] Importantly, after assessing a sample of various commercially available bracket systems, Cash et al[37] found that all bracket slots measured were oversized as compared to manufacturer claims. Likewise, Meling et al[10] suggested a high degree of variation in bracket slot heights, and others have since similarly reported brackets from various manufacturers may be either over or undersized versus manufacturer claims[19,22,25,36,38,39], although heights may still fall within tolerance limits.[19] Interestingly, the bracket slot tops in particular have been found to be oversized while the slot base shows minimal dimensional error, giving the slot of many bracket systems a divergent profile.[25] These dimensional inaccuracies are at least partially due to the manufacturing processes of fixed orthodontic appliances. Specifically, it has been noted bracket slots often have a rough surface with porosities and imperfections as a result of both the manufacturing process and bracket slot polishing process,[18,40] and such imperfections can contribute to dimensional inaccuracies in the slot.[18]

Ultimately, inaccurate fabrication of brackets will alter the amount of play between the slot and the orthodontic archwire, and thereby will affect control of the axial inclination of teeth, thus imprecise fabrication of orthodontic brackets can adversely affect torque control during orthodontic treatment, especially if the bracket slot is oversized as compared to manufacturer claims.[18,24,37,41,42] Indeed, the observation that actual play between wire and slot is often greater than theoretical expectations is partially attributed to variations in bracket slot dimensions.[11,18,22,24] Conversely, literature suggests variations in slot dimensions may lead to only a 2 degree difference in torque expression, which may not be clinically relevant.[38]

1.8.2 Bracket Material

Orthodontic brackets today are available in multiple different construction materials, but SS is generally regarded as the standard option. Specifically, as reviewed in Proffit et al[1], two varieties of SS brackets are available, fabricated through one of two techniques: metal injection molding (MIM) or casting. Most appliances are made via MIM, but the most precision in bracket slot size is achieved by milling the slot of a cast bracket after fabrication. Alternatively, titanium appliances are also available, with advantages of these systems being reduced bonding failures and avoidance of allergies associated with SS. Regarding more esthetic options, non-metallic brackets are also available, including ceramic and plastic options. Plastic brackets are plagued by staining and discoloration, poor dimensional stability, and high friction with metallic wires. To overcome these limitations, plastic brackets generally have a metallic slot, although this does not avoid all the associated challenges with these systems. Ceramic brackets have good esthetic properties and are dimensionally stable, but are brittle so are associated with fractures, and can damage enamel both due to wear if the patient occludes on the bracket and during appliance removal. Importantly, the construction material of a bracket can also influence torque expression.

Specifically, regarding the role of bracket material on torque expression, construction material of brackets can influence torque expression partially due to differences in stiffness and strength of the bracket material contributing to deformation of brackets. For instance, in 1975, Dobrin et al[43] investigated early generations of plastic brackets and noted rectangular wires tended to deform tested brackets. Indeed, the brackets demonstrated an unacceptable amount of deformation and creep when subjected to forces in the physiological range for incisor torque. Improvements have since been made to increase stiffness of plastic appliances to some degree.[18] It follows that when plastic and metallic brackets have been compared more recently, higher torquing moments are consistently generated with metal brackets than plastic brackets, although authors advise plastic brackets can be used for torquing movements provided additional torque is incorporated to offset torque losses with brackets made of these materials today.[44,45] Incorporating a metal slot into plastic brackets has also been shown to sufficiently reinforce these brackets to enable sufficient torque capabilities.[46] Nonetheless, despite these improvements, plastic brackets are still generally not routinely advocated for cases where torquing movements of the teeth are required.[1,38]

Deformation of the bracket is not unique to plastic brackets, however. The literature suggests both elastic and plastic deformation of metallic brackets, including SS and titanium brackets, such as increases in slot height and changes in slot shape, can contribute to increased play, loss of torque expression, as well as reduced torque expression when torque angle is decreasing as compared to increasing when metallic bracket systems are utilized.[21,47–49] For example, Major et al[47] found up to 7.7 degrees of additional torque play may be due to plastic deformation of a metal bracket, depending on the system under investigation, although others find that although play is increased after use of a metallic bracket, the increase may not be statistically or clinically significant.[41] Critically, the threshold of wire twist or torquing moment required to produce deformation ultimately depends on the bracket system under investigation.[48,49] Moreover, the degree of deformation produced in metallic brackets tends to vary depending on the region of the bracket slots assessed.[49]

Comparing ceramic brackets to other alternatives has revealed ceramic brackets generate greater torquing moments[38,50], potentially as a result of greater stiffness of ceramic. For instance, one investigation found nearly two times greater torquing moments with ceramic than polycarbonate brackets, potentially as a result of differences in stiffness between the two materials.[38]

Hardness of the slot can likewise influence torque expression. For instance, Fischer-Brandies et al[11] have shown torque applied to SS slots leads to notching and bending of the slots, and thereby increased play and a reduction in torque expression in the system, largely due to relatively low hardness of brackets as compared to wires. Likewise, fracture resistance can play a role. For instance, regarding ceramic brackets, fracture of the bracket can occur with excessive forces, although fracture resistance is generally considered adequate for clinical use.[51]

1.8.3 Archwire Shape and Dimensions

Archwire dimensions, like slot dimensions, can influence the play between a bracket slot and the archwire, and thus axial inclination control. In general, larger wire sizes are associated with reduced archwire play[52] and higher torquing moments.[53–55] Unfortunately, archwires, like bracket slot dimensions, are often either over or undersized relative to manufacturer claims[11,15–17,19,22,23,25], with the dimensional inaccuracies outside of tolerance limits in some cases.[19] These discrepancies likewise contribute to play between a wire and bracket slot often being found to be greater clinically than expected based on theoretical calculations[11,15–18,22–24], impeding efficient inclination control.

In turn, the shape of the archwire can influence torque efficiency in the pre-adjusted appliance. For instance, Papageorgiou et al[33] compared square and rectangular wires in both 18 mil and 22 mil appliances and found rectangular wires were more efficient in torque expression. Specifically, measured moments generated with rectangular wires were statistically significantly greater than measured moments exerted by square archwires in both 18 and 22 mil appliances, although the effect was exaggerated in 18 mil systems. Likewise, Katsikogianni et al[55] found higher mean torquing moments using rectangular as opposed to square archwires in 18 mil slots.

Edge bevel of an archwire can also influence the play in the wire-bracket system and thus torque expression, where edge bevel refers to the rounding of the corners of a square or rectangular wire. In general, the larger the edge bevel, the greater the play between a slot and archwire.[14–16,23] Critically, edge bevel can show considerable variability, contributing to increased torsional play clinically compared to theoretical calculations.[11,14–17,25,41] Moreover, the four edge bevels of a given archwire often differ from one another and may be irregular[23], thus play may differ depending on the direction of wire rotation. Interestingly, the contribution of edge bevel to deviation angle is also dependent upon the wire material.[14]

1.8.4 Archwire Material

For orthodontic tooth movement, an archwire should generally provide light, continuous forces. In particular, archwires should show good springiness (low stiffness) and a large range so as to provide a light sustained force, and adequate strength to resist permanent deformation.[1,56,57] In some clinical situations, formability and weldability is an advantage so bends and auxiliaries can be incorporated into the wire for detailed control of tooth movement.[1,56,57] Other desirable features include high resilience, or stored energy.[1,56,57] Obtaining a balance of these features for optimal tooth movement is challenging – a wire strong enough to resist distortion may be quite stiff, and thus the force provided may be high initially with rapid force decay.[1]

To meet the demands of orthodontic archwire characteristics at different stages of treatment, archwires are available in numerous materials, each with different mechanical properties. Popular options today include SS, multi-stranded SS, NiTi and TMA. Roughly, SS wires are characterised by high stiffness, low range, high strength, good formability, and low resilience; braided SS wires show low stiffness and good range; NiTi wires have low stiffness, good range, and good resilience; and TMA wires have characteristics somewhat between SS and NiTi, with average stiffness, high resilience, adequate range, and good formability.[1,56,57] In turn, among wires of a given material, mechanical properties of archwires are also affected by wire diameter and shape, length, as well as manufacturer.[1,16,17,52] Specifically, increasing wire diameter will increase strength and stiffness while decreasing range, while increasing length will reduce strength, while increasing springiness and range.[1]

Ultimately, the mechanical characteristics of different materials translate to suggest SS wires are regarded as providing high forces that dissipate rapidly, while NiTi wires provide low relatively constant forces even with large activations.[56,57] TMA wires may be regarded as an ideal choice when forces less than those provided by SS but more than those provided by NiTi are desired.[56,57]

In general, if the degree of torque applied to a wire in a bracket slot is low, the wire is not engaged in the slot and so torque expression or torque moment is not significantly different between different wire materials.[54,58] However, once the wire is engaged, SS is found to have greater torque expression (higher torquing moments) as compared to other alternatives for a given degree of torque applied[22,32,53–55,58], while NiTi wires have the lowest torque expression.[58] In other words, SS wires are shown to express clinically relevant torque at smaller torque angles than NiTi wires of the same dimensions.[22] More specifically, Archambault et al[58] found that at twist angles greater than 24 degrees, SS wires yield 1.5 to 2 times greater torque expression than TMA, and 2.5 to 3 times that of NiTi. Beyond 40 degrees of torque, the situation changes, however. In particular, beyond 40 degrees of torque, differences in torque moment between archwire materials has been found not statistically significantly different[54], or loss of torque expression may occur, especially with stiff wires.[58] These observations are possibly related to the finding that regardless of wire material, deformation can occur in the system when torque is applied, leading to reduced torque expression with use.[11,21,44] Indeed, deformation of the wire may primarily account for the observation that when torque angle is decreasing, torque expression is significantly reduced compared to when torque angle is increasing, as opposed to deformation of the bracket, although both factors contribute.[21]

Importantly, the tendency of SS to produce the greatest torquing moment for a given degree of wire twist is not necessarily an advantage. As previously noted, a physiologically acceptable range of torque lies between 5 and 20 Nmm. Because of the narrow working range and high stiffness of stainless steel, especially as wire sizes increase, only a small range of degree of twist provides physiological torquing moments, thus it is challenging to apply appropriate physiologic torquing moments with stainless steel wires with sufficient accuracy. In other words, the narrow working range of rectangular steel wires makes torque control clinically difficult.[1,15,52] Likewise, the tendency for NiTi to provide low torquing moments with a given degree of twist is not necessarily an advantage – a significant amount of twist may need to be placed in a NiTi wire to apply physiologic torquing moments.[17] Nonetheless, some suggest NiTi wires may be a reasonable choice for applying torque in some situations.[52]

Critically, wire material may affect torque expression to a greater extent than wire size[53]. In addition, it should be noted archwire material does not affect torque expression by influencing play in the system, unlike several factors discussed thus far, but rather influences torque expression due to differences in elastic modulus, or stiffness, between different materials.[22]

1.8.5 Degree of Wire Twist

When orthodontic archwires are engaged to produce third order movements, the torque moment will generally increase as the "degree of torque", "torque angle", or "twist angle" in the wire is increased relative to the bracket slot[12,21,22,27,38,53,54,58,59], although some bracket systems show a loss of torque at high torque angles[21,58], or a plateau in torque at high torque angles.[12] In turn, at low torque angles, often times no torque is generated, depending on the bracket system employed. The rationale for this observation is that the play in the system has not been consumed and the wire is freely rotating within the slot without engaging the walls of the slot in some systems.[21,38]

Given clinically relevant torquing moments fall between 5 and 20 Nmm, it follows that a certain threshold of twist is required to generate adequate moments for orthodontic tooth movement. Considering stainless steel wires as an example, Brauchli et al[38] found a 19 x 25 mil SS wire in the majority of tested brackets with a 22 mil slot required between 20 and 25 degrees of twist to express a clinically significant torquing moment of 10 Nmm.

1.8.6 Direction of Wire Twist

Direction of wire twist may also have an influence on torque expression in contemporary orthodontic appliances for a number of reasons, although this factor has not been intensively investigated. For one, whether torque angle is increasing or decreasing can impact torque expression, with torque expression shown to be substantially less when torque angle is decreasing, likely due to deformation of both the wire and bracket, although wire deformation likely primarily accounts for this observation.[21] Secondly, whether buccal or palatal torque is being applied can influence torque expression in some systems, although torque capacity of most brackets appears to be similar for both directions of torque.[38] Similarly, a thesis on the subject noted some systems generated different torque patterns with a clockwise as compared to a counter clockwise rotation of the wire, especially at higher degrees of wire rotation, when 19 x 25 mil SS archwires were employed in a 22 mil slot.[60] These observations were speculated to relate to various aspects of the bracket design, including asymmetry in bracket slot depth when comparing gingival and incisal slot walls.

1.8.7 Bracket Prescriptions

Orthodontic brackets are available in a suite of different prescriptions, including different torque prescriptions, referring to the built-in inclination of the bracket slot. These prescriptions may be built into the face or the base of the bracket, known as torque-in-face or torque-in-base, respectively. The latter is more common today, with one commonly cited benefit being improved slot alignment.[8,61] Different appliance prescriptions claim to offer superior esthetics, resistance to unwanted tooth movement, improved occlusal outcomes, among other purported benefits.[8,61]

In general, larger torque prescriptions are thought to generate greater moments and thus increased torquing of teeth. Specifically, common prescriptions used today include Roth and Mclaughlin, Bennett, Trevisi (MBT) prescriptions, each differing in degrees of prescribed torque. Considering the maxillary central incisor, the MBT prescription incorporates 17 degrees of palatal root torque (+17 degrees of torque) while Roth includes 12 degrees of palatal root torque (+12 degrees of torque) for the same tooth, with the increased torque in the MBT prescription advocated to avoid a retroclined appearance of upper incisors.[8] In turn, high and low torque options are also often available with many bracket systems. Notably, the pre-programmed torques in each of the common prescriptions are generally greater than reported ideal inclinations of the teeth at the end of treatment. The rationale is partially to overcome issues of archwire play, among other factors.[8] Indeed, when comparing inclination of teeth after treatment to the prescription incorporated in the bracket used, one investigation found the final tooth positions did not match the incorporated prescription[62], perhaps due to play in the system.

The clinical impact of bracket prescription on torque expression is somewhat controversial, however. For instance, when considering *in vivo* investigations, Moesi et al[63] compared outcomes achieved with MBT and Roth prescription brackets and found bracket prescription had no significant effect on the subjective outcome of treatment. Likewise, others have found bracket prescription has no clinical influence on treatment outcomes.[64] Indeed, it should be noted slop may completely negate any incorporated torque in a pre-adjusted appliance in some instances making the prescription clinically irrelevant, as evidenced by the observation by Joch et al[19] that 19 x 25 wires in a 22 mil slot may have as much as 11.7 degrees of slop, which is nearly equivalent to the 12 degrees of maxillary central incisor torque pre-programmed into an appliance adhering to the Roth prescription. Others have similarly reported pre-programmed torque values may be irrelevant, considering the wide range of archwire play before torque is generated.[38,50] When considering a clinical situation, it also must be noted slop will not only be observed at a tooth of interest, but also at adjacent anchorage teeth, thus further increasing play between archwire and bracket[44,45], which could additionally influence the clinical relevance of an incorporated prescription. Finally, it must be noted manufacturing processes not only contribute to variability in slot dimension, but also variability in torque prescription and bracket base, which can further impact torque expression.[18]

In contrast, when comparing high and low torque prescription brackets *in vitro*, Sifakakis et al[31] found maximum moments generated with a given wire twist with high torque brackets were statistically significantly higher than those generated with low torque prescriptions using the same dimension archwire and degree of wire twist, both for 18 and 22 mil appliance. It follows that some authors have concluded high-torque prescriptions should generally be selected to try to account for lack of expression of prescribed torque values.[18]

1.8.8 Dental Anatomy and Bracket Placement

Yet another factor that plays a critical role in torque expression with pre-adjusted appliances is tooth anatomy and bracket placement, given these factors can influence the inclination of the slot relative to the archwire. Essentially, dental anatomy and bracket placement can alter the degree of twist of the archwire relative to the bracket slot, thus influencing torque expression.

Specifically, regarding the influence of dental anatomy, significant differences exist in the contours of the labial surfaces of the dentition, including labial surface inclination, such that the facial surface can change the orientation of the bracket slot. In other words, due to variable tooth morphology, engagement of the same archwire in the same bracket on different teeth will result in different root torque.[65–67] Critically, the magnitude of variation in tooth morphology is so great so as to negate standard torque prescriptions in some cases.[65,66] Further in support of the influence of tooth morphology on torque expression, finite element analysis has demonstrated that anatomy plays a critical role in strain developed in the PDL with torque application.[68] Interestingly, central incisors may be the least variable in contour when considering the maxillary anterior dentition.[69]

Accurate bracket placement can likewise have a strong influence on torque expression, with different positions of the same bracket on the same tooth producing important differences in root torque.[67] More specifically, Meyer and Nelson[70] found a vertical error in bracket placement of 3mm can alter the torque of a premolar by as much as 15 degrees. Likewise, finite element analysis has revealed buccolingual displacement is significantly influenced by vertical bracket positioning.[68] Unfortunately, bracket placement often shows considerable variability and thus variability in torque expression due to bracket placement is expected. For example, Balut et al[71] found considerable discrepancies in bracket placement, including both vertical and angular discrepancies, which can be attributable to operator skill, tooth structure, and tooth position. Importantly, the effect of bracket placement on torque expression can occur even if brackets are placed within a clinically acceptable position.[67]

For similar reasons, positions of brackets on adjacent teeth and positions of adjacent teeth themselves can be anticipated to influence torque applied to a given tooth clinically, given a single straight wire is often employed throughout each dental arch in a clinical scenario. Specifically, variables associated with adjacent teeth can be hypothesized to impact inclination of the archwire relative to the bracket slot on a tooth of interest, and thus torque expression at a tooth of interest.

1.8.9 Ligation Methods

Contemporary appliances can employ several ligation methods – strategies to hold the archwire within the bracket slot. Roughly, conventional systems utilize steel or elastomeric ligatures to hold the wire in the bracket slot, while self-ligating (SL) systems use rigid clips, spring clips, or retaining springs to hold the wire in the bracket slot.

Specifically, considering conventional systems, the traditional ligation technique involves the use of 8 to 14 mil stainless steel ligature wire engaged over the wire-bracket complex rigidly retaining the wire in the slot (Figure 10A). In contrast, elastomeric ligatures consist of circular synthetic elastomers made either through injection molding or cut from previously processed elastomeric tubing (Figure 10B). Elastomeric options are touted to provide continuous, gentle forces, long lasting archwire seating, resistance to water sorption, and shape memory.[72]

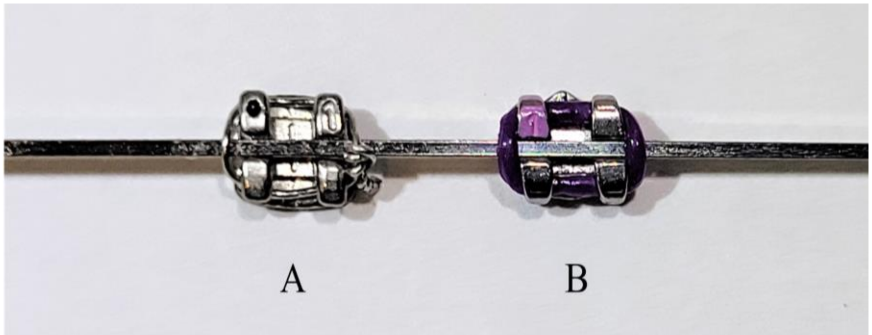


Figure 10: Conventional twin brackets mounted on a 19 x 25 mil stainless steel wire, ligated with a stainless steel ligature tie (A), and an elastomeric ligature (B)

Regarding SL systems, these appliances were introduced as early as the mid-1930s in an effort to reduce chair time associated with steel ligation methods[38,73], and consist of a door mechanism on the bracket that can be opened for insertion and removal of archwires and closed to retain the wire in the slot (Figure 11). As compared to conventional systems, SL brackets are suggested to enhance patient comfort, improve oral hygiene, reduce treatment time, and enhance patient acceptance, among others.[73] Indeed, Turnbull and Birnie[74] demonstrated SL methods statistically significantly reduced ligation time and overall chair time compared to conventional techniques.

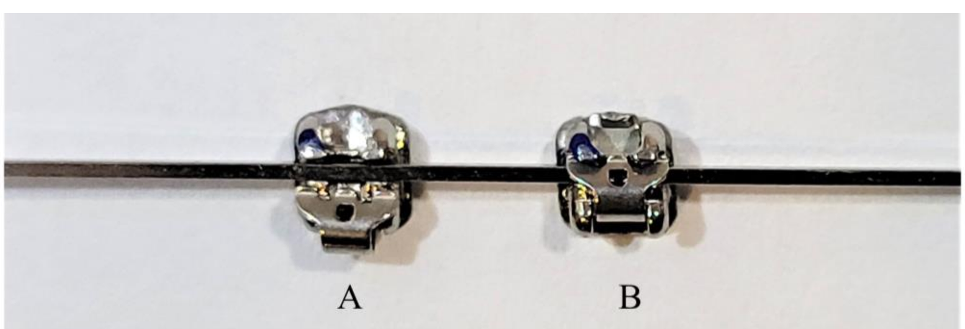


Figure 11: Self-ligating brackets with the door mechanism in the open (A) and closed (B) position mounted on a 19 x 25 mil stainless steel wire

Critically, two types of SL systems are available on the market today: active self-ligating (ASL) and passive self-ligating (PSL) systems. Comparing ASL and PSL systems, ASL systems use a clip-type closing mechanism that will actively press the archwire into the slot, while PSL systems use a closing mechanism that essentially acts to transform the bracket into a tube (Figure 12). Reported benefits of the ASL systems include enhanced control of tooth movement due to the archwire seating mechanism of the active clip, while PSL systems are promoted to reduce friction between the archwire and the bracket slot.[38,73]

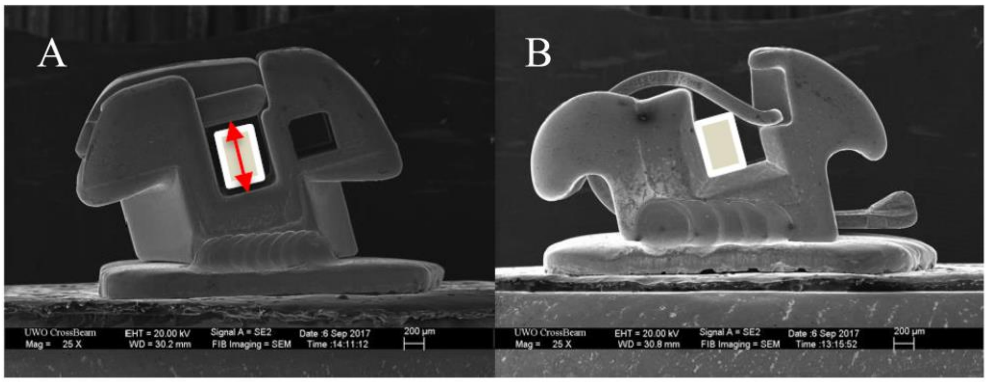


Figure 12: Scanning electron microscopy images of a passive (A) and active (B) SL bracket. The former highlights the door-type mechanism providing freedom to the wire within the slot, while the latter demonstrates the active clip mechanism seating the wire within the bracket slot. Image used with permission from a previous thesis completed by Greene et al[75]©

Critically, ligation method may influence torque expression in contemporary appliances. For example, regarding conventional systems, it is often suggested steel ligation techniques are associated with superior torque expression as compared to elastomeric ligatures. The rationale is multifactorial. For one, it has been suggested steel ligation could reinforce the bracket against deformation during torque expression. Indeed, Al Fakir et al[59] found steel ligatures are associated with greater torque expression and less bracket deformation as compared to elastomeric ligatures with conventional brackets. Alternatively, it has been suggested steel ligatures hold the wire more rigidly in the slot than elastomeric options, pushing the wire into the base of slot, and thereby may reduce play in the system and increase torque expression, even in the face of oversized slots.[18,24] Similarly, elastomeric ligatures deliver a restraining effect that delivers a small amount of torque, even if archwire play is not eliminated, but the amount of this torque contribution is small and clinical significance is questionable.[52] Moreover, elastomeric ligatures show rapid force degradation (50-70% force loss within 24 hours of use), less force exertion, and permanent deformation, thus some have suggested the use of these ligatures should be limited to initial alignment and levelling phases of treatment, with limited application for rotational and torque corrections.[72]

In turn, the effect of steel versus elastomeric ligatures in conventional systems may depend on wire size. For instance, Hirai et al[54] found that after eliminating play, torque with elastomeric ligatures is less than with steel ligatures if wires are undersized, but elastomeric ligatures do not lower torque if full sized wires are being utilized.

Regarding SL systems, significant controversy exists in the literature as to the clinical significance of ASL versus PSL designs on torque control. ASL techniques that push the wire into the slot have been suggested to reduce rotation of the wire within the slot, allowing torque expression even before the wire sufficiently rotates within the slot to engage the side of the slot. Support for this theory comes from observations that ASL systems have less play than other systems, and in some cases, less play than expected based on geometric calculations.[12,53] Likewise, others have found archwire play is often larger than anticipated, especially with regards to PSL systems[24], and a study investigating friction in ASL and PSL systems found the two ligation strategies first engaged different sized wires.[76] It follows that, ASL systems are often advertised to offer superior third order movement control. In support of these claims, several authors have compared both 22 and 18 mil ASL and PSL systems and determined ASL systems are more effective in torque expression than PSL systems.[12,24,55] For instance, Badawi et al[12] assessed 19 x 25 SS wires in 22 mil slot active and passive SL systems and found ASL brackets required 15 to 31 degrees of torsion to generate a clinically effective torque of 5-20 Nmm, while 22.5 to 34.5 degrees of torsion was required to generate equivalent torque in passive systems.

In contrast, others have found only minor differences between ASL and PSL systems, inferior performance of ASL systems, or no differences between the two systems at all. Specifically, Major et al[21] found torque play between the different self-ligation bracket styles was not clinically significantly different, as well as noted torquing moments were not statistically significantly different with up to 24 degrees of twist (corresponding to 34Nmm of torque and less) when considering 19 x 25 SS wires in 22 mil slots. Moreover, in some instances, ASL systems have been found to have lower torquing moments than other systems[53,60], and PSL systems have been found to express clinically effective torque at lower torque angles than other systems according to some authors.[77] Some of these findings may be due to some ASL systems demonstrating greater tendency to bracket slot deformation under torsion as compared to other systems.[47,48] It follows that some have concluded the influence of an active versus a passive SL method on torque expression is of little clinical significance.[38] Furthermore, some *in vivo* investigations have not demonstrated the purported third order control with these systems, with no significant difference in torque expression between the two ligation types.[78]

More conclusively, regarding active versus passive SL systems, a systematic review of the literature published in 2010 found ASL brackets had an engagement angle of about 7.5 degrees, compared to PSL systems, which demonstrated an engagement angle of approximately 14 degrees using 19 x 25 steel wires in a 22 mil slot, with engagement angles generally greater than theoretical values. Moreover, the authors concluded clinically effective torque was achieved in active systems with 15-31 degrees of torsion and in the passive systems with 23-35 degrees of torsion.[20] Contrariwise, more recently, another systematic review concluded only minor differences in torque expression occur between active and passive SL brackets[27], adding to the controversy of the influence of active versus passive SL techniques.

Comparing conventional to SL systems, again significant controversy exists in the literature: some investigations have found superior torque capabilities in the SL systems[55], while others report both SL and conventional systems can be equally efficient in torque delivery[60,79], and others still have shown some evidence conventional brackets generate larger maximum torque compared to other systems.[53] Some ASL systems may be inferior to both conventional and PSL systems in torque expression.[60] Similarly, some find less play in conventional systems as compared to SL systems, suggesting superior inclination control with the former.[24] More decisively, a recent systematic review determined conventionally ligated systems presented with higher torque expression compared to SL systems[27], and as such, some have suggested SL brackets are not recommended if torque control is a focus of treatment.[24]

Ultimately, in interpreting results comparing ligation methods, it must be noted that a large proportion of investigations have utilized large dimension rectangular wires, with few investigations considering other wire dimensions. Whether the picture changes when smaller wire dimensions are employed thus remains to be fully established, as it is possible the spring clip of some active systems allows earlier torque expression than is possible with passive systems. Thus far, the literature available comparing torque expression using active, passive, and conventional ligation methods in multiple wire sizes is limited and suggests conflicting results.[53,55] Likewise, direction of rotation may affect active versus passive self-ligation systems differently, given some have demonstrated ASL systems generated significantly different torque when subject to clockwise versus counter clockwise wire rotation[60], although others have found little effect of direction of wire rotation regardless of bracket system.[38]

1.8.10 Other Factors of Note

Finally, although not always at the forefront of the literature investigating torque expression in orthodontics, some additional factors may influence torque expression in everyday clinical practice. For example, after assessing new, used, and recycled stainless steel brackets, it has been noted that nickel content is decreased in retrieved and recycled specimens, suggesting ion release occurs *in vivo*[80], perhaps secondary to a synergistic action of temperature variations with bacterial activity and bacterial by-products in the oral environment.[18] This ion release could impact properties of the bracket and thus torque expression during *in vivo* use of bracket systems. Likewise, during orthodontic treatment, it is commonplace that complex combinations of forces, moments, and couples are applied to a tooth, with first, second, and third order movement occurring simultaneously. These complex force systems can impact torque expression. For example, some authors have noted second-order misalignments during torque application result in statistically significant differences in torque expression.[81] Likewise, it has been demonstrated that application of a second-order couple will establish a small third order couple and thereby increase torque exertion for small angles of twist (below 22 degrees), such that effective torque can be generated at twist angles below those predicted based on torsional play.[82] However, once torsional play is eliminated, the effect of a second order couple on torque expression is minimal.[82] Conversely, increasing torque angle can reduce the critical contact angle required for second order tooth movements.[83]

1.9 *In-Vitro* Methodology for Studying Torque to Date

Various methods are available by which torque expression has been assessed *in vitro*. For illustration, one systematic review of the literature identified eleven studies of torque expression meeting the author's inclusion criteria, and among those eleven investigations, eight different devices were used to quantify torque expression.[20]

More generally, one method of studying torque commonly reported in the literature utilizes the orthodontic measurement and simulation system (OMSS). This system was thoroughly described by Drescher et al[84] in 1991 and consists of two force-moment sensors capable of measuring forces and moments in all three dimensions, each mounted on a positioning table possessing six degrees of freedom. Notably, the assembly is built into a temperature controlled chamber[84], and the apparatus can ensure measurements are being made in only a single plane of space by automatic adjustments of the system through a load-sensor feedback system.[20] A typical setup using this device involves a dental model with levelled and aligned arches to which brackets are bonded using a full-sized steel archwire such that archwires can be inserted into the brackets passively. A sensor of the OMSS replaces the tooth of interest in the arch, with a bracket bonded directly to the sensor.[31–33,44,45,50,55] Critically, both absolute and simulation measurements can be recorded with the device. Specifically, absolute measurements of forces and moments are obtained with the OMSS by recording forces and moments generated by an appliance as the positioning tables are moved, whereas simulation measurements are obtained by determining the forces and moments generated by an appliance, then calculating the movements of teeth, and finally moving the positioning tables accordingly to record the movement path of teeth and the force systems developed.[84]

A second method of torque assessment commonly noted in the literature centers on a novel apparatus developed in 2008 by Badawi et al[12] which several authors have since employed in modified forms.[21,48,58,59,81] Specifically, the apparatus developed by Badawi et al[12] consists of a torsion system with a digital inclinometer used to measure torsional rotation of a wire (angle of torsion), a wire support substructure to hold the wire and ensure rotation occurs along the wire long axis, an alignment assembly including turntables that can adjust the bracket position to ensure perfect alignment of slot and archwire and thereby ensure torque is measured while other forces and moments are kept to zero, and worm-gear to rotate the archwire segment. The bracket is mounted on a force/torque transducer which measures forces and moments in all three planes of space. Torque values can then be determined at the load cell as in Badawi et al[12], or by modifying the methods to include data transformation, values at the bracket slot itself can be determined.[21,48] Yet alternatively, torque can be measured at an estimated center of resistance (ie 10mm from the bracket slot). Differences in torque magnitudes recorded in different studies could be partially attributable to differences in where torque is being measured, and this must be kept in mind when interpreting data in the literature.[21]

Alternatively, torque expression can be measured using different styles of lathes [41,44,46], among other apparatuses.[22,24,38,46,51,52,54,60,77] For instance, Franco et al[77] utilized a wire torsion device associated with a universal testing machine, while Brauchli et al[38] welded or bonded brackets to a screwhead that was then mounted on the frame of a hexapod while wires were fixed within a drill chuck mounted on a force/moment sensor that was in turn mounted on a rotary disk. In turn, Dalstra et al[24] bonded brackets to an aluminum bar and subsequently employed a Force System Identification machine to determine forces and moments in the system. Perhaps most recently, Young et al[60] developed an apparatus utilizing an Instron E10000 to measure torquing moments generated with wire rotation.

Interestingly, torque values may be found to be lower for experiments using the OMSS than for other experimental designs, possibly because of adjacent teeth in the model providing increased play in the system as compared to other designs.[44,45]

Notably, many of these torque measuring devices developed to date require extensive machining of components, expensive equipment, and some demonstrate considerable measurement noise, possibly due to large load cell capacities relative to torque values being measured, representing some significant limitations in the literature to date.

1.10 Summary of Gaps and Conflicts in the Literature

Evidently, literature available examining torque expression in different bracket systems is somewhat limited. In particular, limited research is available comparing ASL, PSL, and conventionally ligated brackets, especially using wires other than large dimensioned SS wires in different directions of torsion. Moreover, the available literature is wrought with many controversial conclusions. For instance, some indicate ligation method has little impact on torque expression while others suggest ligation technique may significantly influence torque expression. Likewise, some have suggested torque expression is similar regardless of whether buccal or palatal root torque is applied, while others suggest that for ASL systems, direction of torsion may significantly impact torque expression. It is imperative that the knowledge base surrounding torque expression is expanded to include data examining the influence of wire size in different ligation systems along with direction of wire rotation, given that in clinical practice multiple wire sizes are employed and different directions of torque application are commonplace. Likewise, additional data is required to help resolve conflicting data currently available. Ultimately, a prudent clinician should possess a full understanding of the appliances employed in everyday clinical practice so as to efficiently and effectively treat patients and bridging the aforementioned gaps will facilitate such.

Given the limitations in methods of studying torque to date, consideration for development of a compact, low-cost, accurate, and precise measurement alterative for studying torque is warranted. Namely, modern technology, including the advent of rapid and affordable three-dimensional printing technology, has recently offered a potentially new avenue for development of an effective and affordable apparatus design.

Chapter 2

2 Purpose and Hypotheses

2.1 Purpose of the Current Investigation

The aim of the current investigation was to both develop a compact, table-top apparatus utilizing three-dimensional (3D) printing technologies for accurately and precisely measuring torque expression by orthodontic appliances, and ultimately to compare torque expression using various wire sizes in ASL, PSL, and conventionally ligated orthodontic brackets, in the equivalent of both buccal and palatal root torque. This represented a follow-up investigation to a previous study examining friction generated by conventional, ASL, and PSL systems in different wire sizes, and which demonstrated significant differences in friction generated between different ligation methods with different wire sizes.[76] Ultimately, such an investigation would enhance understanding of the influence of wire size, ligation method, and direction of torque application on torque expression in commonly available orthodontic appliances. It was anticipated that this information would help guide orthodontists in choosing the appropriate bracket systems and wire sizes for each individual patient, depending on the torquing needs of the case.

2.2 Hypotheses

1. ASL bracket systems generate significantly greater torquing moments than PSL and conventional systems for a given degree of rotation in all wire sizes

2. Larger wire sizes generate greater torquing moments than smaller wire sizes for a given degree of twist in all bracket systems

3. Torque direction (buccal or palatal root torque) will influence torquing moments generated with ASL bracket systems, more than PSL or conventional systems, in all wire sizes

4. ASL bracket systems demonstrate a reduced engagement angle as compared to PSL or conventional bracket systems in all wire sizes

Chapter 3

3 Materials and Methods

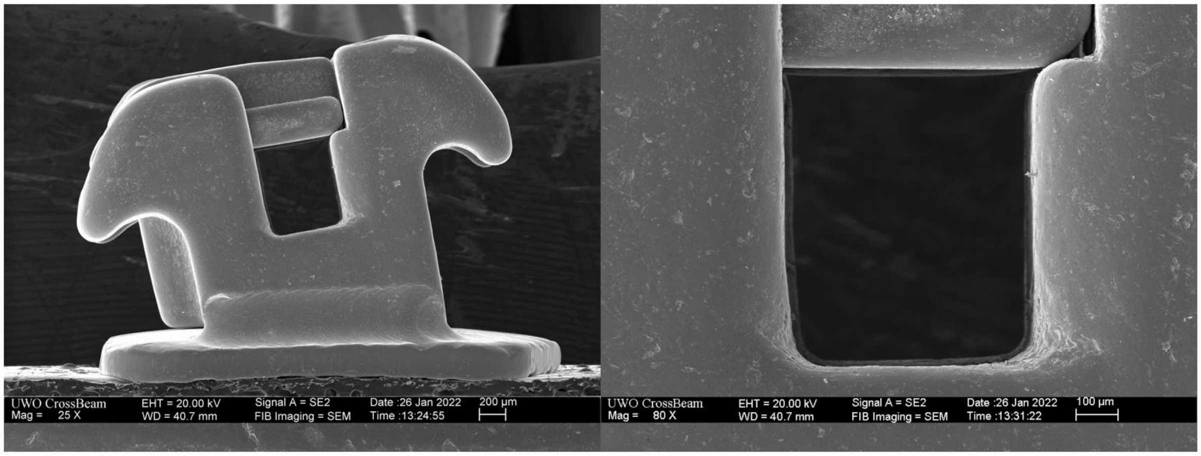
3.1 Brackets of Interest

Three 22 mil ASL bracket systems were tested in torque expression, along with a 22 mil PSL bracket system, and a 22 mil conventional twin bracket system with an elastomeric ligature as a control. Specifically, ASL systems tested comprised of Empower®2 Metal Interactive SL brackets, Speed System™ brackets, and 3M™ Victory Series™ ASL brackets, while the PSL system investigated was the Damon™ Q2 PSL system. A conventional 3M™ Victory Series™ Twin bracket using a new grey elastomeric ligature for ligation was used as a control. These systems were selected for analysis according to bracket popularity, bracket availability, and use in previous literature. All tested brackets comprised upper right central incisor brackets. The prescription employed was the most common prescription available for each system. Differences in torque prescription between systems did not influence results due to the manner in which torque expression was evaluated: brackets were mounted for testing so as to zero any incorporated prescriptions in the bracket. A summary of features of each bracket system examined are shown in Table 1. Scanning electron microscopy images of each bracket system were obtained and are shown in Figure 13.

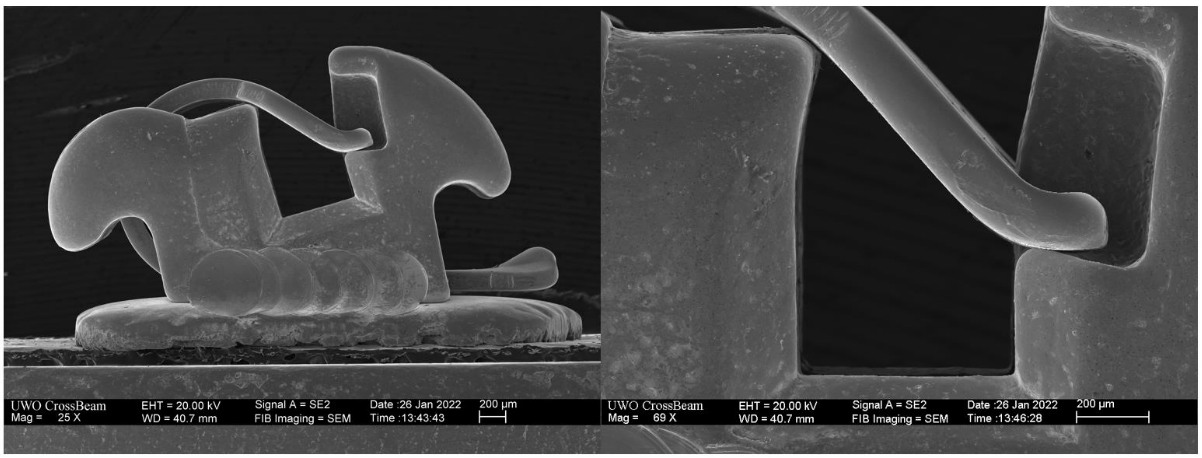
Ligation Method	Bracket System	Test Group	Manufacturer	Prescription	Item Number	Lot Number(s)
PSL	Damon™ Q2	P-Dmn	Ormco™ (Brea, CA, USA)	Damon™	491-8860	07217405N
ASL	Empower®2 Metal Interactive	A-Emp	American Orthodontics (Sheboygan, WI, USA)	MBT	485-1117	034084
	Speed System™	A-Spd	Strite Industries (Cambridge, ON, Canada)	MBT	22UR1+17HR	081321 102221
	3M™ Victory Series™ ASL	A-Vic	3M™ Unitek™ Orthodontic Products (Monrovia, CA, USA)	Roth	025-302	LD1UU
Conventional (Elastomeric)	3M™ Victory Series™ Twin	C-Vic	3M™ Unitek™ Orthodontic Products (Monrovia, CA, USA)	MBT	017-876	LS9VE
	Elastomeric ligature	-	American Orthodontics (Sheboygan, WI, USA)	-	854-660	N37341

Table 1: Summary of features of the bracket systems of interest

A



B



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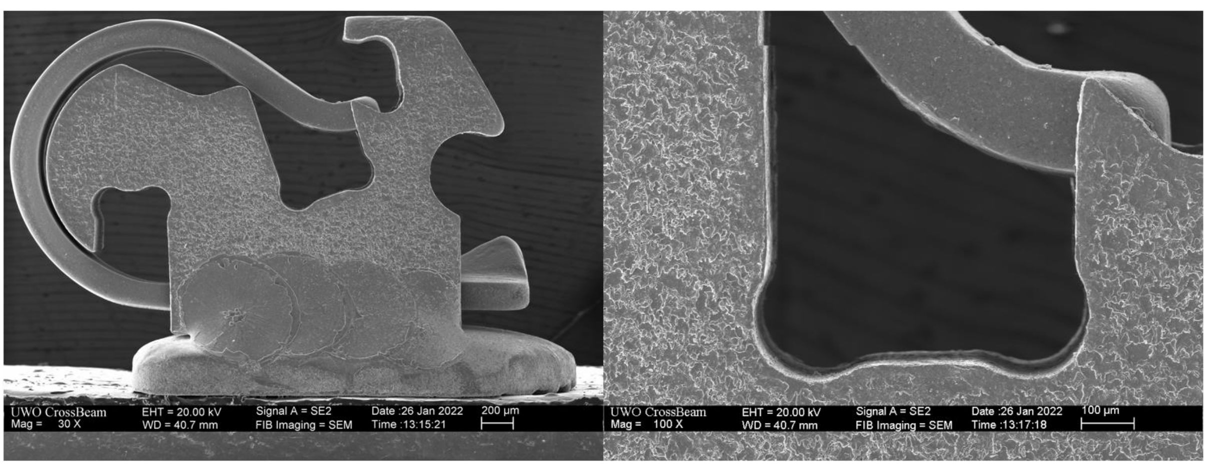

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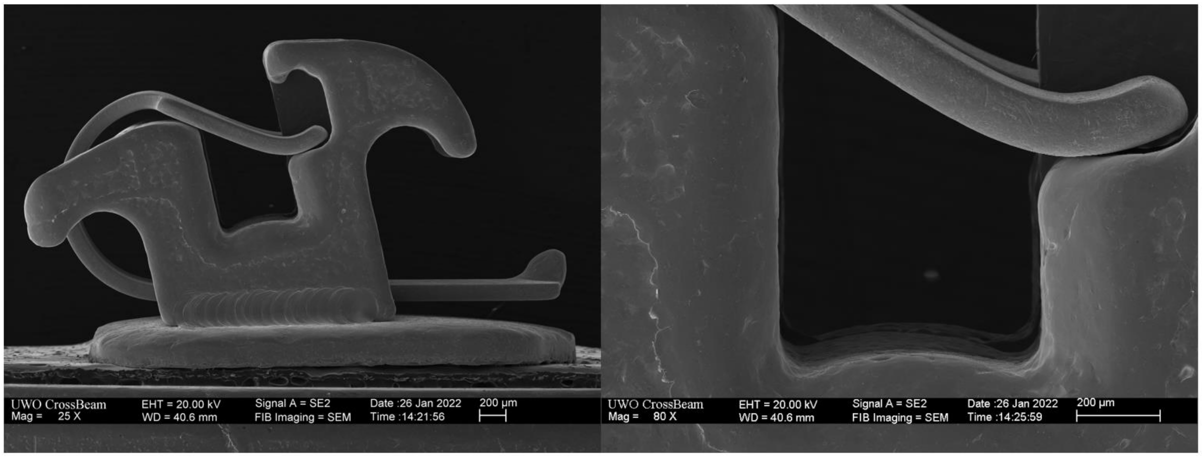
D



E

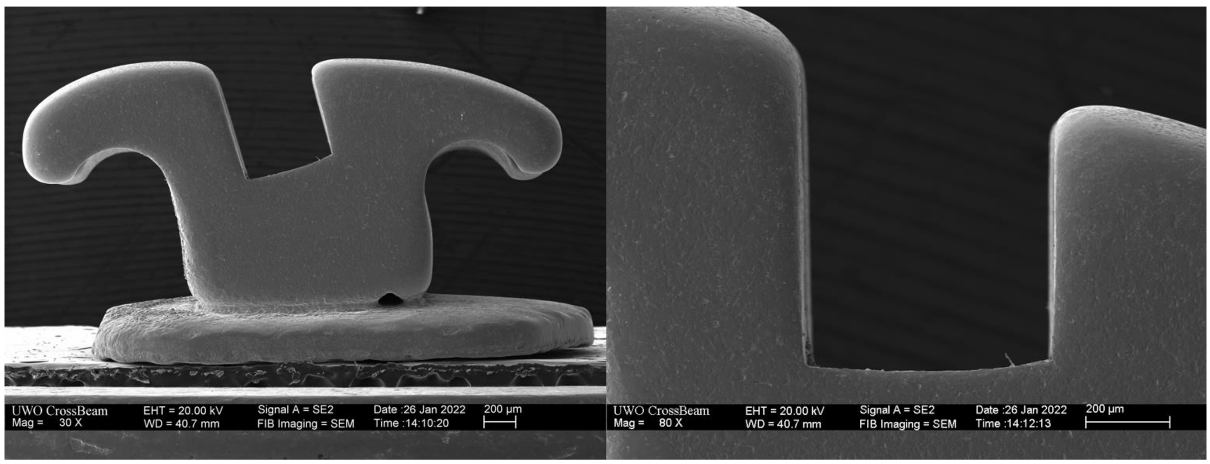


Figure 13: Scanning electron microscopy images of each bracket system of interest, with (A) showing P-Dmn, (B) A-Emp, (C) A-Spd, (D) A-Vic, (E) C-Vic. Leftward images represent the entire bracket with the gingival aspect oriented rightward, rightward images represent a magnified image of the bracket slot itself. Magnification and references scales are shown on each individual image. Various measurements obtained within the bracket slot are found in Appendices A-E

3.2 Wires of Interest

Four SS wire sizes were examined: 16 x 22, 17 x 25, 19 x 25, and 20 x 20 mil SS wires, each sourced from G & H Orthodontics (Franklin, IN, USA) in 14-inch straight lengths. The wire sizes tested were not necessarily representative of the wire sequence recommended for use by bracket manufacturers as many different wire sequences are traditionally used with different bracket systems. Rather, wires of these sizes were selected due to their generally common use in clinical practice, frequent recommendation for use with ASL systems, as well as due to findings from a previous investigation examining friction in various bracket systems with different wire sizes.[76] Characteristics of the employed wires are shown in Table 2. Notably, wires as measured with a digital caliper consistently measured 1 mil smaller in both dimensions than manufacturer reported specifications. Noting nominal dimensions differed from manufacturer size designations was consistent with the available literature.[11,15–17,19,22,23,25]

Wire size	Manufacturer	Item Number	Lot number
16 x 22 mil	G & H Orthodontics (Franklin, IN, USA)	STSS1622	907652
17 x 25 mil		STSS1725	85986
19 x 25 mil		STSS1925	926631
20 x 20 mil		STSS2020	777623

Table 2: Summary of features of the investigated orthodontic wires

3.3 Apparatus: Mounting Jig

A custom mounting jig as described by Young et al[60] and Greene et al[75,76] was adapted for the current investigation to mount tested brackets onto newly custom fabricated hexagonal SS transfer pins (Figure 14A) for testing in torque expression. The jig, shown in Figure 14B, consisted of two rectangular aluminum poles on which clamps were attached, connected to one another via an aluminum base. The base of the jig contained a slot into which a single hexagonal SS transfer pin could be accurately positioned and secured with a screw to the jig. The clamps accepted a 21.5 x 28 mil SS wire used for bracket mounting. Such a large dimension wire segment should have nearly completely filled the slot of a 22 mil bracket to allow repeatable mounting of the tested brackets to a transfer pin while eliminating pre-programmed prescriptions in the appliances. Brackets were secured to the transfer pin by assembling the jig and transfer pin securely together, inserting a 21.5 x 28 mil steel wire into the jig clamps, ligating a bracket to the wire, and bonding the bracket to the transfer pin.

To ensure repeatable, precise positioning of the brackets along the transfer pins, a crimpable stop was placed on the wire such that lateral aspect of the brackets sat flush against the stop during the bonding procedure. This stop was positioned laterally to the midpoint of the transfer pin so as to allow centering of the bracket midpoint along the transfer pin. The precise positioning of the stop was adjusted for different bracket systems to account for differences in widths of different bracket designs, thereby allowing centering of each bracket system midpoint along the hexagonal transfer pin. The same positioning of the crimpable stop was used for each round of bonding of a given bracket system to ensure repeatable positioning of the bracket for each replicate.

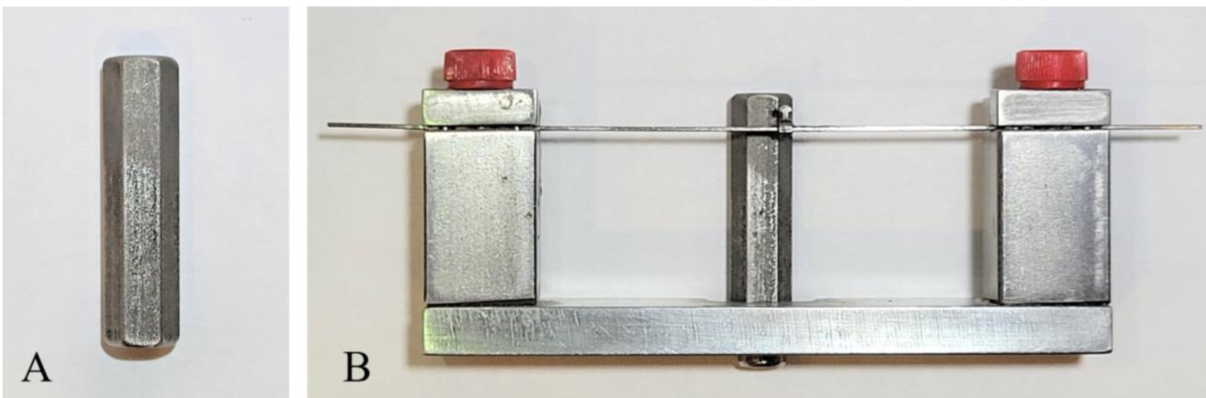


Figure 14: Prepared hexagonal transfer pin (A), and assembled mounting jig including a 21.5 x 28 mil SS wire with crimpable stop and installed hexagonal transfer pin (B)

3.3.1 Bracket Mounting Protocol

Tested brackets were mounted onto the SS transfer pins using the mounting jig outlined above. Specifically, protocol for mounting involved micro-etching the bonding surfaces of the transfer pins using 50 micron aluminum oxide (item#15301, Lot#L0BWZ and 43186, Danville Materials, Carlsbad, CA, USA), application of Assure® Plus All Surface Light Cure Bonding Primer (item# PLUS, Lot# 215151, Reliance® Orthodontic Products, Itasca, IL, USA) to the transfer pin, air thinning of the primer, and application of light cure GoTo™ adhesive to the bracket base and transfer pin (item# GTP, Lot#214169, Reliance®

Orthodontic Products, Itasca, IL, USA). The bracket was aligned relative to the transfer pin using the mounting jig and the adhesive was light cured after removal of any excess. Throughout the bonding procedure, all materials were handled with gloves to avoid introduction of contaminants.

Importantly, for every wire size of interest, ten (10) brackets of each system were mounted for testing both with the gingival aspect of the bracket oriented downward relative to the mounting jig, and with the gingival aspect directed upward relative to the jig (Figure 15). This allowed testing to simulate both buccal root torque (movement of the tooth root toward the patient's cheek or lip) and palatal root torque (movement of the tooth root toward the patient's palate or tongue), respectively.

Figure 15: Prepared hexagonal transfer pin and bonded bracket with the gingival aspect of the bracket oriented upwards. This bonding arrangement simulated palatal root torque during testing. A downward orientation of the gingival aspect of the bracket simulated buccal root torque

3.4 Apparatus: Custom Torque Assembly

A custom, table-top torque assembly was fabricated to evaluate torquing moments generated by the various bracket-wire combinations. The basic function of the device involved controlled rotation of individual brackets of interest with respect to a fixed segment of the wire sizes of interest. The assembly was fabricated with a combination of aluminum and 3D printed plastic components and was designed such that each wire size of interest could be centered within the bracket slot and coaxial with the axis of rotation of a stepper motor and torque measuring component. Considering some elements of the apparatus consisted of plastic, finite element analysis of the apparatus design was conducted with a maximum expected torque of 250 Nmm to ensure stress and deformation would not significantly impact measured torque or result in long-term fatigue or distortion.

Roughly, the apparatus consisted of a stepper motor; the torque-measuring fixture, which involved a load cell and load cell mount mounted on a rotational base; a rotational arm onto which the hexagonal transfer pin and mounted bracket was affixed; and a base fixture with custom aluminum wire clamps that firmly held the wire size of interest in position relative to the other components (Figure 16).

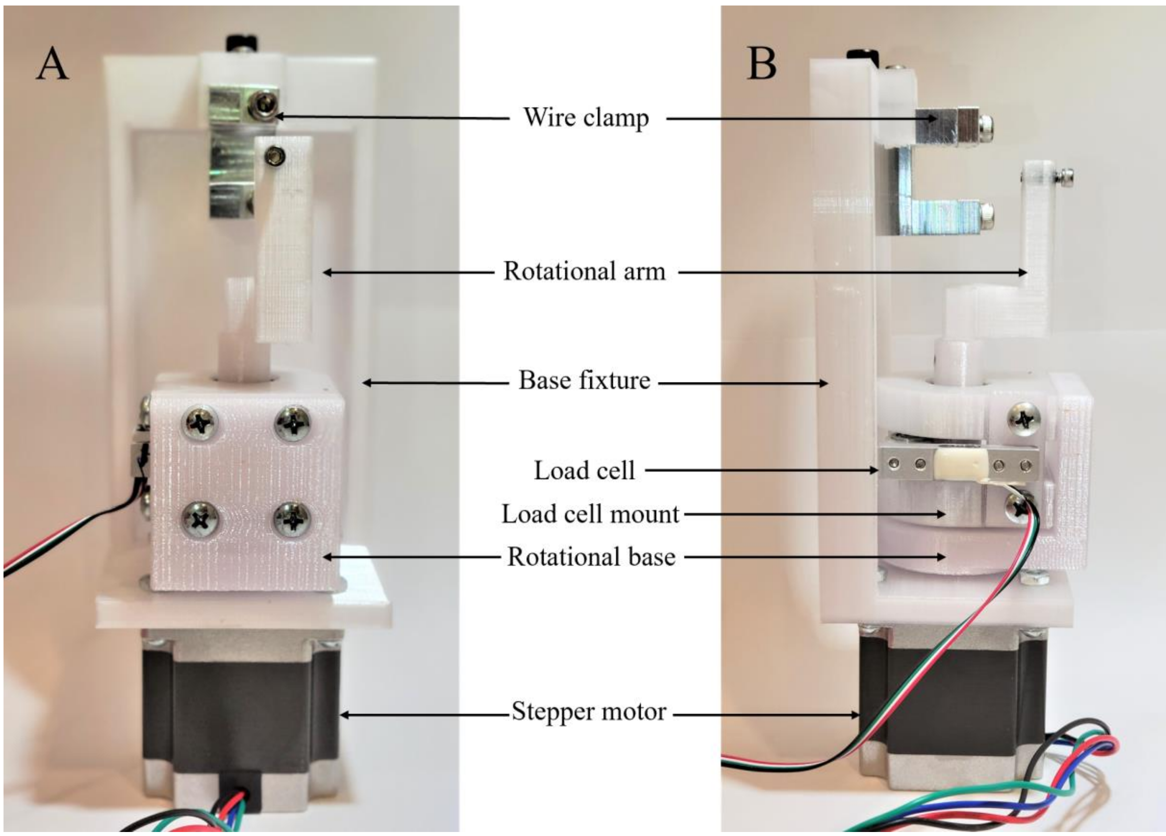


Figure 16: Custom torque assembly as viewed front the front (A) and side (B) perspectives. Labels highlight the main components of the assembly: the stepper motor, torque-measuring fixture (load cell and load cell mount mounted on a rotational base), rotational arm, and base fixture with wire clamps

Specifically, the base fixture of the device was an L-shaped component that was 3D printed using polyethylene terephthalate glycol (PETG) using a commercial 3D printer (Dremel 3D45, Robert Bosch Tool Corporation, Mount Prospect, Il, USA). To the superior aspect of the base fixture, wire clamps fabricated of aluminum were secured so as to hold the wire size of interest stationary throughout testing. These wire clamps were custom fabricated for each wire size of interest to account for variations in the central axis of each wire type and were fastened to the base fixture using a dowel and screw. The dimensions of the clamp that accepted the wire size of interest were fabricated to be 1 mil smaller in both dimensions than manufacturer reported wire dimensions to accommodate the tendency for wire sizes to measure toward the lower tolerance limits specified by manufacturers. The span of wire between the clamps was precisely 15 mm to simulate the span of wire used to torque an upper right maxillary central incisor in clinical practice. Aluminum as opposed to PETG was selected as the material of choice for clamp fabrication to limit wear and fatigue within the clamp fixture throughout testing.

To the underside of the base fixture, a Nema 23 bipolar stepper motor (Part Number: 23HS22-2804S, OSMTec, Jiangbei District, Ningbo, China) was mounted using nut and bolt fixtures. The rotational base was 3D printed using the same PETG filament and printer as for the base fixture and was press-fit to the stepper motor shaft. A 2mm thick ring on the undersurface of the rotational base held the rotational components away from the base fixture to allow clearance during testing.

In turn, a 780 gm Wheatstone bridge load cell (RB-Phi-117, Robot Shop Inc, Mirabel, Quebec, Canada) was mounted onto a 3D printed load-cell mount, and this complex was mounted onto the rotational base using screws. The load-cell mount was aligned such that the axis of rotation of the load cell was centered with the center axis of the wire size of interest and stepper motor. The rotational arm was printed using the same materials and printer as for the other components. This component slid into a central shaft of the load cell mount and was fixed in place with a set screw. This arm was designed so as to accept a single hexagonal transfer pin and its mounted brackets and thereby held the bracket at the center of the wire size of interest, with an axis of rotation aligned with the center of the wire size of interest. Torque was transmitted from the rotational arm to the load cell through the load cell mount central shaft, supported within two ball-bearing assemblies (17mm ID, 35mm OD, Model 6202.2ZR.L38, FAG Bearings). The shaft was connected to the load cell through a 3D-printed PETG connecting rod, attached with two pairs of flanged miniature bearings (3.175 mm ID, 9.525 mm OD, Model RB-SCT-1220, Robot Shop Inc, Mirabel, Quebec, Canada) to reduce friction. The torque-measuring fixture was designed to measure over the range of ±200 Nmm, with a safe overload range of ±240 Nmm and precision of ±0.1 Nmm. An isometric CAD view of the torque-measuring fixture (specifically the load-cell mount and load cell) is shown in Figure 17 with transparency added to highlight the innerworkings of this component.

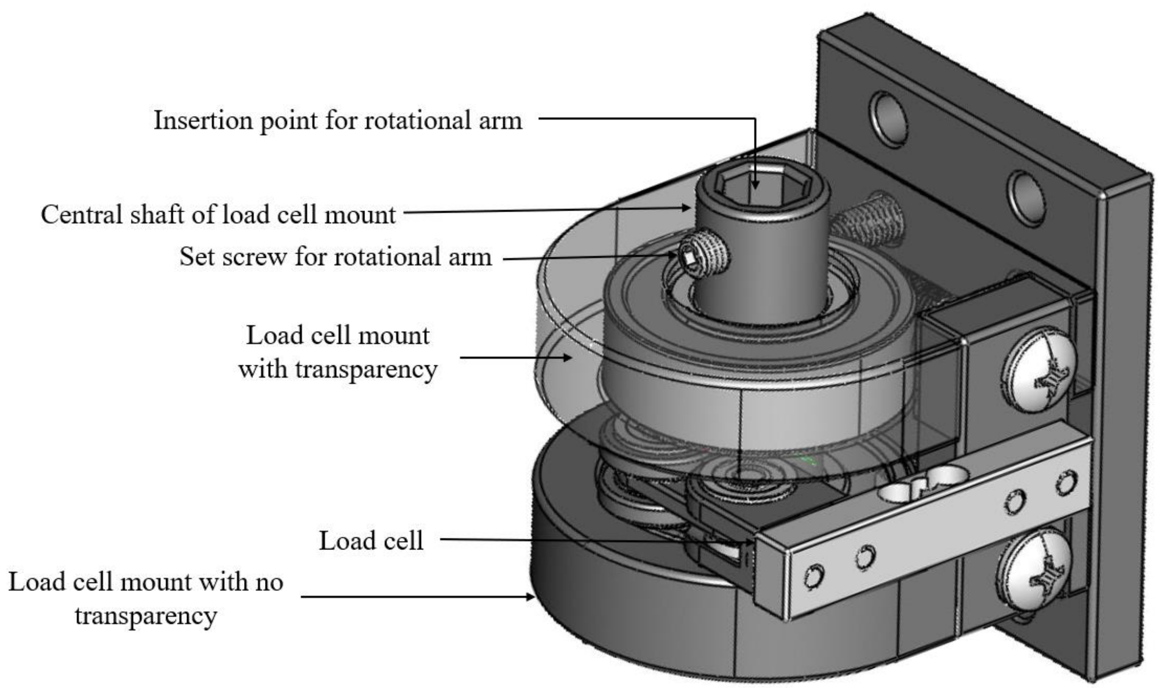


Figure 17: Isometric CAD views of the load-cell mount and load cell with transparency added to highlight the innerworkings of this component. Labels highlight some of the various components of the torque-measuring fixture

A removable hand-held alignment jig was likewise 3D printed and was designed such that it could be pressed against the base fixture and rotational base to hold the assembly at a zero-angle position to ensure reproducibility of the starting position.

The device was controlled using a microprocessor (Arduino® Uno board) and Python™ programming via a personal computer. Specifically, the target position in degrees and speed of rotation in degrees per second was specified via a custom Python™ graphical user interface (GUI). The output from the Wheatstone bridge load cell was amplified and digitized by a load-cell amplifier (HX711, Avia Semiconductor, Xiamen, P.R. China) and the digital output was transmitted to the microprocessor (Arduino® Uno) for analysis.

3.5 Torque Testing

To initiate a test, once brackets were bonded to transfer pins using the protocol outlined above and the wire size of interest inserted into the wire clamps on the assembly base fixture, a bracket mounted on a hexagonal transfer pin was installed onto the rotational arm of the device and secured to the wire size of interest. A repeatable zero position of the assembly was established with the custom fabricated hand-held alignment jig. Before initiation of each test, the load cell underwent a “tare” protocol with no load applied; this “no-load” torque value was automatically subtracted from subsequent measurements.

The target position in degrees and speed of rotation in degrees per second was then selected, and a test cycle initiated electronically via a personal computer and the custom Python™ graphical interface (GUI). Once the command was input, the stepper motor subsequently initiated rotation, resulting in rotation of the bracket of interest around the wire size of interest. Recorded torque values were sampled from the load cell at increments of 0.11 degrees. Specifications for each test were a 45-degree rotation at a rate of 1 degree per second, converted to the nearest step per second, beginning at a zero-degree angle. For each test, the rotational arm rotated counterclockwise from the zero position, as viewed from above the custom torque assembly. The resulting data was automatically populated to a Microsoft® Excel® (Microsoft 365 MSO (16.0.13801.21002) 64-bit; Microsoft Corporation, Redmond, WA) document specifying the time of the test in milliseconds, target position, current position, and torquing moments in Nmm.

As noted, for each wire size of interest, brackets of each system were tested in different torquing directions by bonding 10 brackets of each system to transfer pins oriented with the gingival aspect either down or up relative to the transfer pin and mounting jig assembly. This allowed comparison of torquing moments generated with the equivalent of buccal and palatal root torque, respectively (Figure 18). Test groups were distinguished according to whether testing simulated buccal or palatal root torque by indicating a "B" or "P" at the end of the test group name (i.e. test group C-Vic(B) represented a Victory Series Twin bracket undergoing a simulation of Buccal Root Torque).

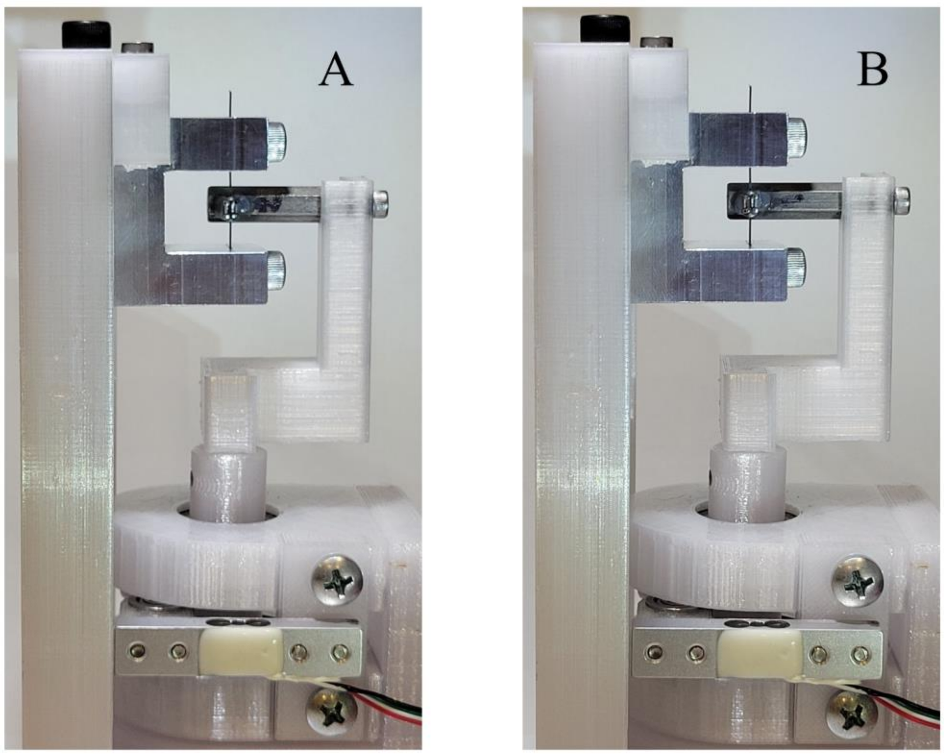


Figure 18: Brackets mounted in the custom torque assembly with the gingival oriented downward relative to the mounting jig to simulate buccal root torque (A), and the gingival oriented upward relative to the mounting jig to simulate palatal root torque (B)

A fresh bracket and wire segment was used for each replicate, and in addition, for conventionally ligated systems, a fresh recently manufactured elastomeric ligature was utilized for each replicate (see Table 1). Tests were all performed at room temperature, and the apparatus was allowed to cool 10 minutes between each 10 replicates to avoid heating of components influencing readings.

3.6 Validation and Calibration of the Custom Torque Assembly

The load cell was calibrated before first use utilizing multiple known weights (each 4.69 grams(g)) in groups of 10 (equivalent to 46.9g) applied sequentially at a known radial distance (28mm) from the axis of rotation of the load cell using a custom calibration wheel mounted to the load cell and its load cell mount. The weight of the calibration weights was verified using a Mettler Toledo Milligram Scale (Columbus, OH, USA). Expected Nmm measurements were calculated based on the weights applied and the diameter from which the weights were suspended from the axis of rotation, and the resultant output from the load cell combined with these expected torques in Nmm were then used to calibrate the device.

Throughout testing, maintenance of calibration was assessed using known weights of 200 and 500g suspended from the rotational arm of the apparatus and comparing the resultant torque readings to readings obtained immediately following initial calibration. Specifically, torquing moments generated with these known weights were recorded before and after each testing group and compared to initial readings of 54.89Nmm and 132Nmm for the 200g and 500g weight respectively. If drift in calibration was detected, a correction factor could be applied to the resultant data if needed. However, measurements remained consistent throughout testing, with 200g weights producing torque measurements with a range of 2Nmm and 500g weights producing measurements with a range of 3.9Nmm. These ranges were deemed adequately precise given clinically significant torquing moments measure a minimum of 5Nmm. As such, no correction factor was applied to the resultant data.

For validation of the custom torque assembly prior to testing, the rotational arm was rigidly affixed to the base fixture and one degree of rotation was applied in both directions from the zero-degree starting position. This validation design would allow quantification of any deformation or flex in the device design.

Specifically, a rigid hexagonal bar fabricated of Chrome Vanadium Steel was clamped into the base fixture of the device. After a zero position of the assembly was established using the alignment jig and the tare protocol completed, a prepared hexagonal transfer pin was inserted into the rotational arm of the device. Assure® Plus All Surface Light Cure Bonding Primer (item# PLUS, Lot# 215151, Reliance® Orthodontic Products, Itasca, IL, USA) was applied to both the transfer pin and hexagonal bar and flowable composite resin (Transbond™ Supreme LV Low Viscosity Light Cure Adhesive, Item#712-046, Lot#NC36419, 3M™ Unitek™ Orthodontic Products, Monrovia, CA, USA) was applied to both surfaces to connect them. This material was then light cured from all dimensions as per manufacturer specifications, thereby rigidly connecting the rotational arm and base fixture.

Once the rigid design was established, one degree of rotation was applied to the device in both a clockwise and counterclockwise direction from a zero starting point and torque moments generated recorded. Based on the resulting measurements, it was determined that stiffness in the system was 45Nmm per degree. In other words, for every 45 Nmm applied to the device, deformation was 1 degree. This is likely a conservative determination as it is likely the adhesive allowed some flexibility in the validation setup itself.

3.7 Data Analysis

3.7.1 Torquing moments

Mean torquing moments were determined for each bracket-wire combination in both torquing direction simulations for every 0.11 degrees of rotation, and these means were plotted as torque-rotation curves (torque in Nmm per degree of twist) for analysis. Likewise, descriptive statistics, including mean torquing moments and standard deviations, were calculated for each bracket-wire combination at the nearest approximation to 15-degree intervals (15, 30, and 45 degrees). After confirming a normal distribution of the data, torquing moment data at these intervals were evaluated with statistical software

(IBM® SPSS® Statistics 27.0; SPSS, Inc., Chicago, IL) using Two-Way analysis of variance (ANOVA) with Bonferroni correction for multiple comparisons to assess the effect of both independent variables (bracket system and wire size) on mean torquing moments generated at each 15-degree increments. Statistical significance was set at $P<0.05$. The Two-Way ANOVA revealed that for each 15-degree increment there was a statistically significant interaction between the effects of bracket system and wire size [15 degrees ($F(27,360)=37.80$, $P<.001$); 30 degrees ($F(27,360)=52.79$, $P<.001$); 45 degrees ($F(27,356)=37.19$, $P<.001$)]. Therefore, One-Way ANOVAs with Bonferroni correction for multiple comparisons were conducted to uncover simple main effects of both bracket system and wire size.

Notably, for analysis, if the adhesive connecting a bracket to its mounting pin was found to have failed during testing, then all data subsequent to the adhesive failure was excluded from analysis as these failures did not represent behaviour of the bracket but rather of the adhesive, which was not under analysis in this investigation. If SL bracket doors visibly opened during testing, the data was likewise eliminated from analysis as this could represent error in the experimental method, such as failure to completely close the doors at the initiation of a test. However, if it was noted the doors deformed during testing (i.e. were pushed laterally or bent outward during testing) this data was maintained for analysis as this represents true *in vitro* behaviour of the bracket. These criteria led to exclusion of four brackets at the 45 degree increment only. Specifically, these exclusions left sample sizes of n=10 for all bracket groups at all increments for both 16 x 22 and 20 x 20 mil wires. For 17 x 25 mil wires, n=10 for all bracket groups except A-Spd(B) at 45 degrees, which had a sample size of n=9. For 19 x 25 mil wires, n=10 for all bracket groups except P=Dmn(P) and A-Spd(B) at 45 degrees which had a sample size of n=9 and n=8 respectively.

3.7.2 Engagement angles

To evaluate the engagement angle for each bracket-wire combination, a mean intercept and standard deviation was determined for each bracket system-wire size combination using computer coding, where the intercept was defined as the angle where a line fit to the linear portion of the torque-rotation curve passed through zero torque (Figure 19). Resultant mean engagement angles were again evaluated using Two-Way ANOVA with Bonferroni correction for multiple comparisons to assess the effect of both independent variables (bracket system and wire size) on mean engagement angles. Again, Statistical significance was set at $P<0.05$, and the Two-Way ANOVA revealed there was a statistically significant interaction between the effects of bracket system and wire size ($F(27,351)=6.81$, $P<.001$). Again, One-Way ANOVAs with Bonferroni correction for multiple comparisons were conducted to uncover simple main effects. Sample sizes were 10 for each group (n=10), except for both A-Vic groups and the A-Emp(P) group with 16 x 22 mil wires which each had sample sizes of n=6 and n=9 respectively due to some replicates never adequately engaging this wire size.

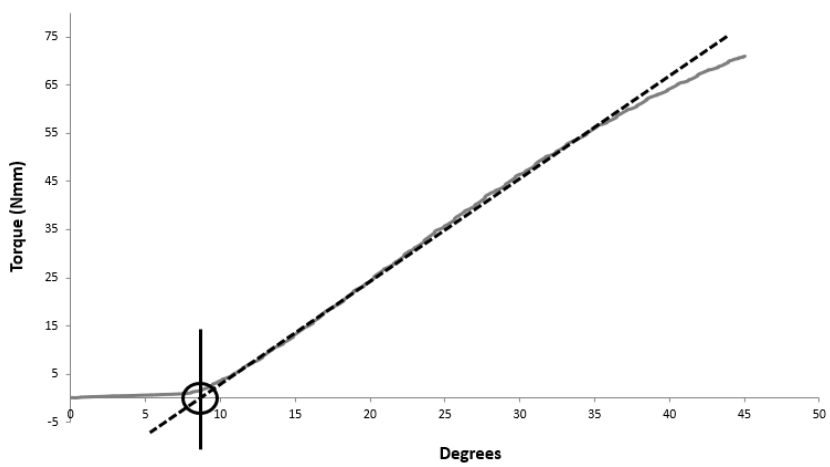


Figure 19: To determine engagement angle, a line was fit to the linear portion of the mean torque-rotation curve by computer code and the angle at which this line passed through zero was assigned as the engagement angle

Chapter 4

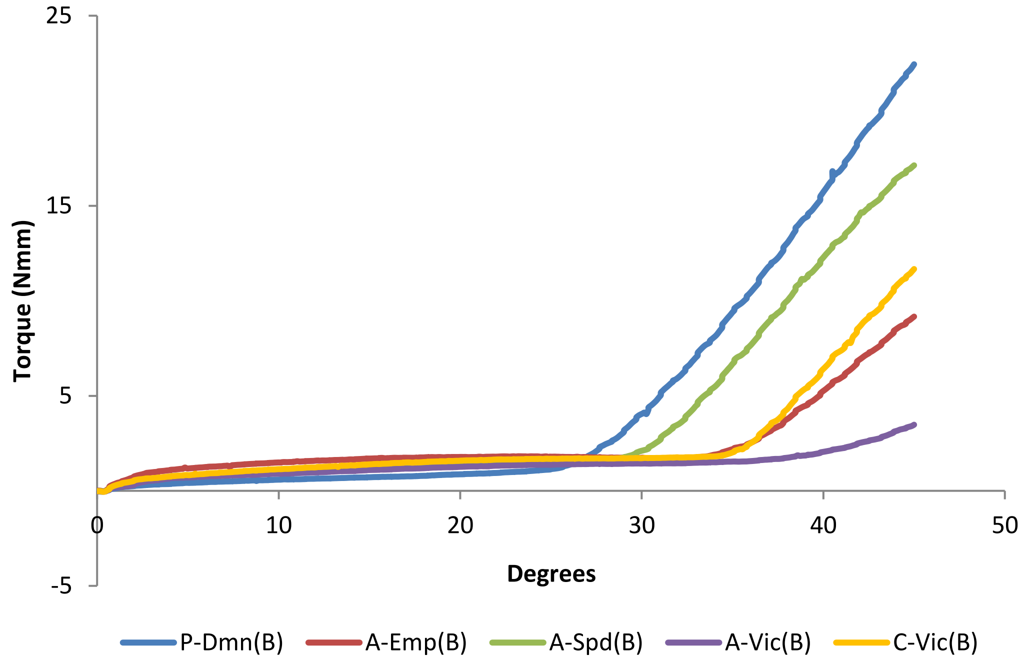
4 Results

4.1 Mean Torquing Moments with Progressive Rotation

Mean torquing moments in Nmm determined for every 0.11 degree increment of rotation from 0 through 45 degrees for each bracket-wire combination in both directions of rotation (buccal and palatal root torque simulations) are plotted in graphical format as torque-rotation curves in Figures 20-23. Mean torquing moments for each bracket-wire combination at the nearest approximation to 3-degree intervals of rotation are included in Appendices F-I.

In general, with increasing rotation, torquing moments increased for each bracket-wire combination in both buccal and palatal root torque simulations, after engagement of the wire within the bracket slot. For 16 x 22 and 17 x 25 mil wires, the relationship between torque and degree of rotation was generally linear after engagement of the wire within the slot, regardless of bracket group. For 19 x 25 and 20 x 20 mil wires however, beyond approximately 35 degrees of rotation, the linear trend between torque and degree of rotation was not maintained – the curve began to reduce in slope at high degrees of rotation. Moreover, for both A-Spd groups and these two wire sizes, a peak torquing moment emerged (approximately 35 and 30 degrees respectively), and in simulations of buccal root torque with these wire sizes (A-Spd(B) groups), a rapid decline in torquing moments was ultimately observed as rotation continued to progress.

A



B

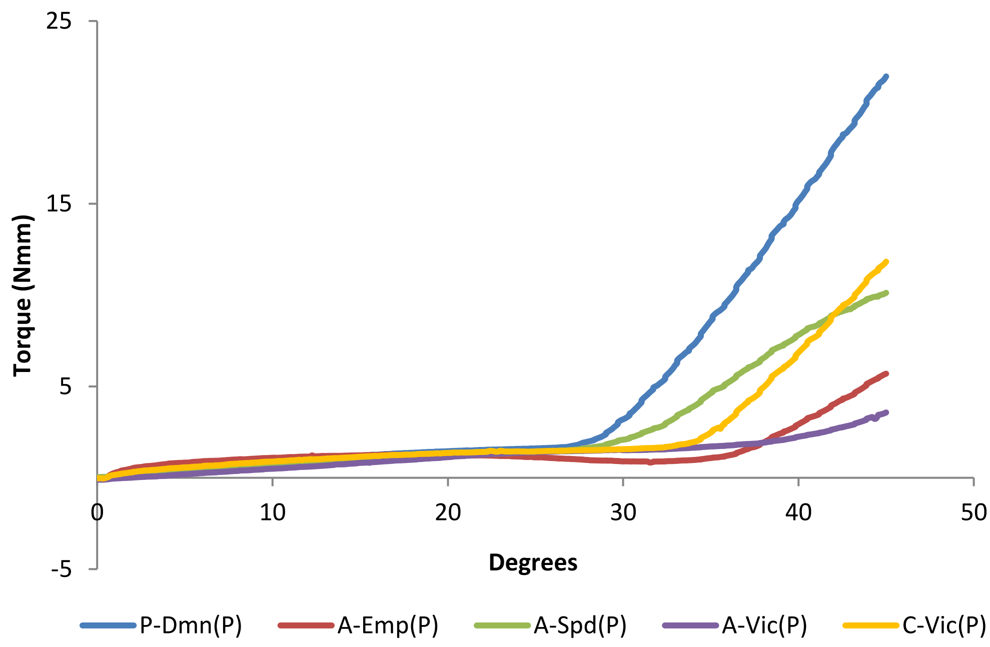
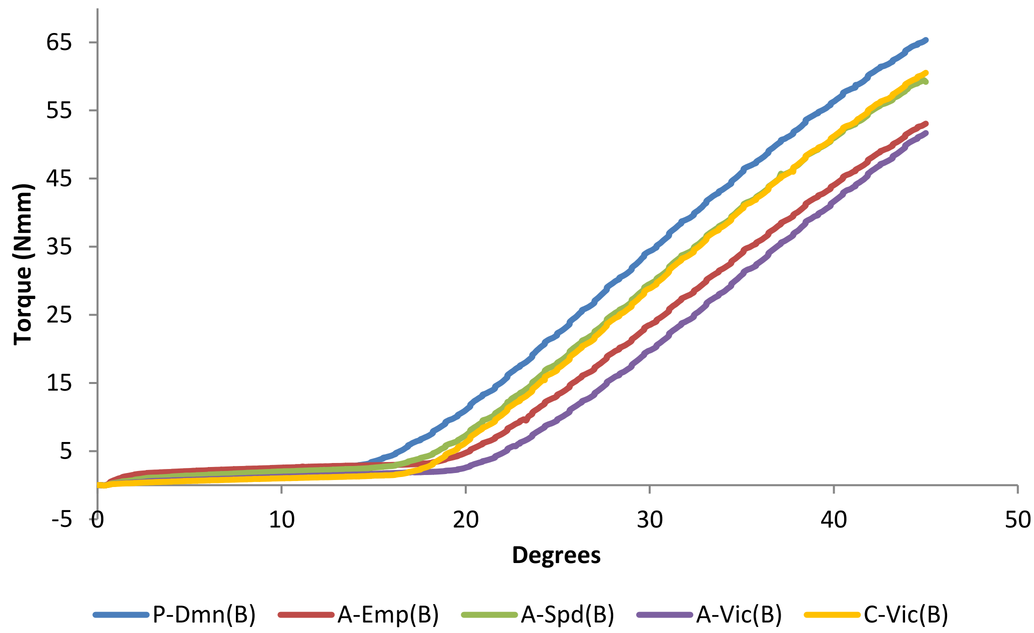


Figure 20: Mean torquing moments measured for each bracket system and 16 x 22 mil SS wires, simulating both buccal (A) and palatal root torque (B)

A



B

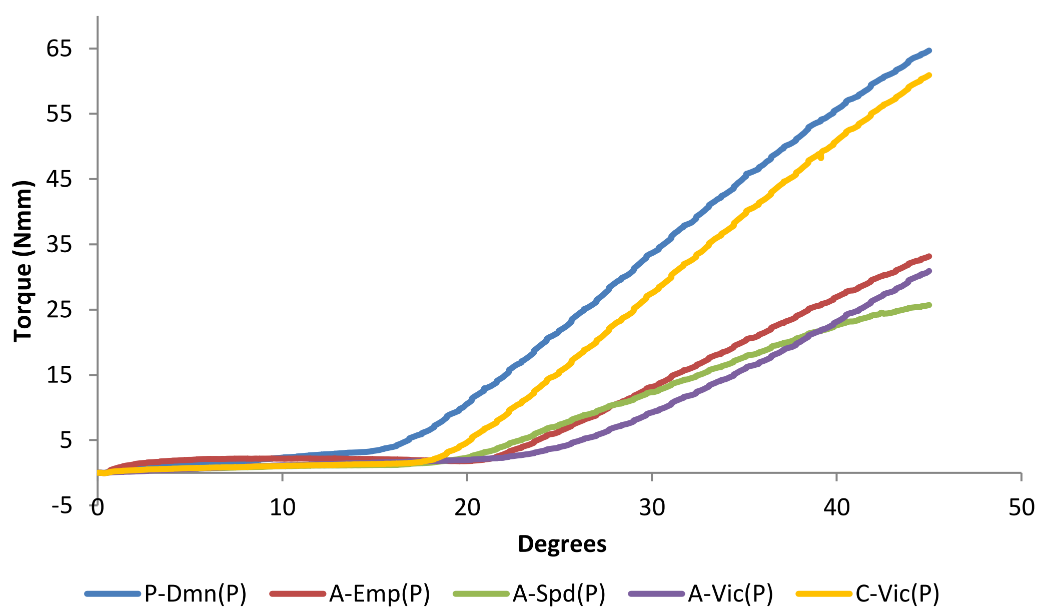
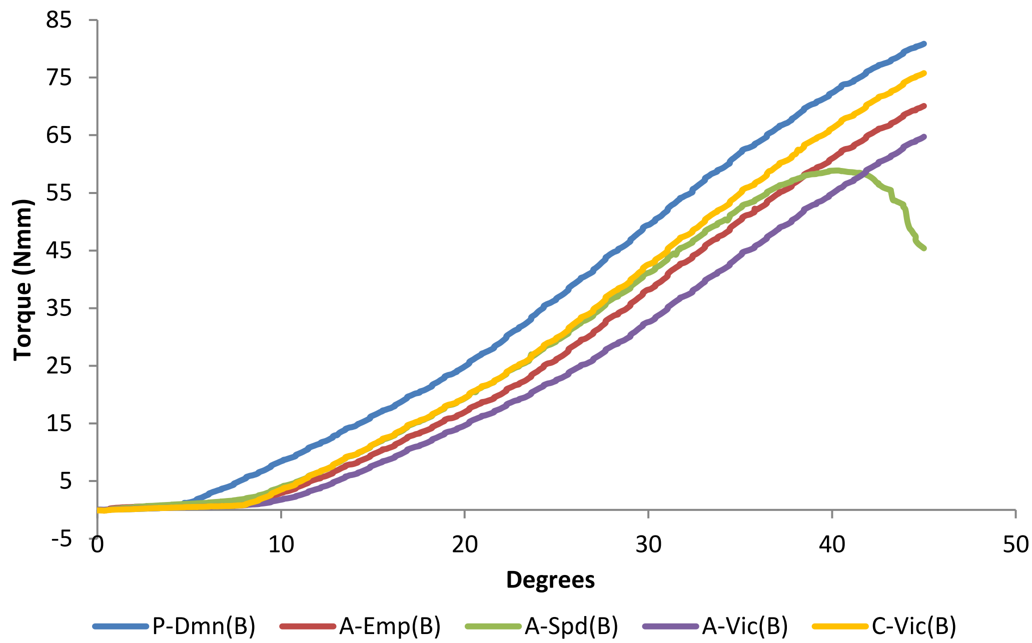


Figure 21: Mean torquing moments measured for each bracket system and 17 x 25 mil SS wires, simulating both buccal (A) and palatal root torque (B)

A



B

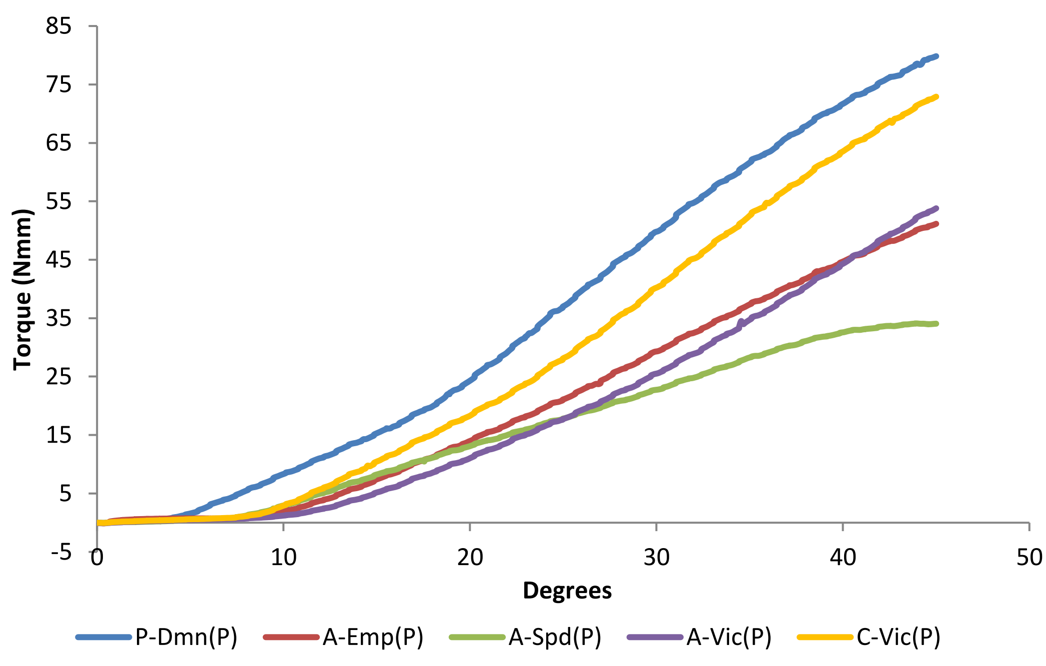
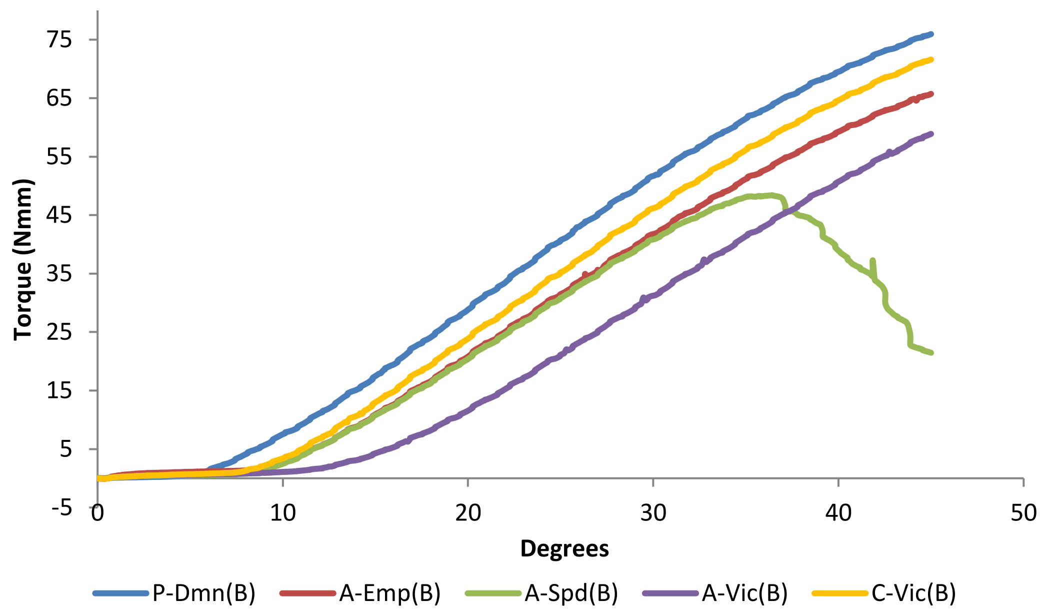


Figure 22: Mean torquing moments measured for each bracket system and 19 x 25 mil SS wires, simulating both buccal (A) and palatal root torque (B)

A



B

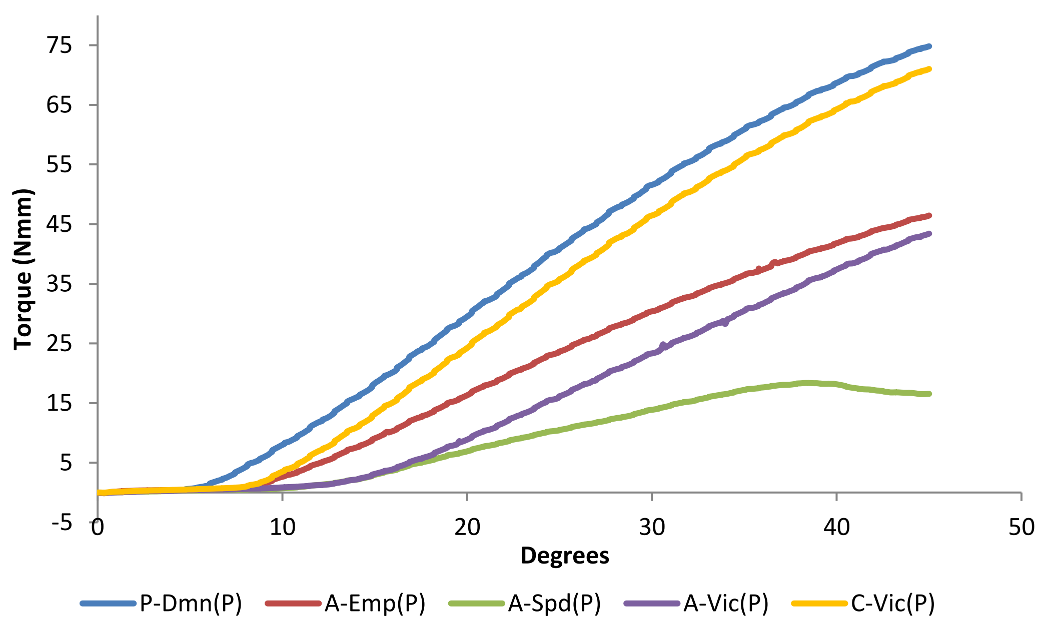


Figure 23: Mean torquing moments measured for each bracket system and 20 x 20 mil SS wires, simulating both buccal (A) and palatal root torque (B)

4.1.1 Comparing Bracket Systems

Comparing mean torquing moments generated by different bracket groups at a given 15-degree increment of rotation within a given wire size revealed a statistically significant difference between bracket groups for all comparisons. Associated means (±SD) along with significance results are highlighted in both Table 3 and Figures 24-26, and associated P-values are included in Table 4.

More specifically, at 15 degrees of rotation, although some significant differences in mean torquing moments were detected between bracket groups with a given wire size, many demonstrated no significant differences, especially among the two smallest wire sizes (16 x 22 and 17 x 25 mil wires). Moreover, no clear pattern emerged as to which brackets generated significantly different mean moments at this rotational interval, especially with the smallest two wire sizes.

As rotation progressed through 30 to 45 degrees, a pattern emerged whereby PSL (P-Dmn) and conventionally ligated (C-Vic) bracket groups tended to demonstrate significantly greater mean torquing moments as compared to the ASL groups, with some exceptions.

In particular, at 30 degrees of rotation with all four wire sizes, both P-Dmn groups (P-Dmn(B) and P-Dmn(P)) consistently generated significantly greater mean torquing moments than all other investigated bracket systems ($P<0.05$). Second largest mean torquing moments at this rotational interval were generated by the C-Vic groups, and in some cases, mean moments generated with these systems were not significantly different from buccal root torque simulations with A-Spd (A-Spd(B)) and A-Emp (A-Emp(B)) groups. Specifically, for 17 x 25 and 19 x 25 mil wires, the A-Spd(B) group showed no significant difference in mean torquing moments compared to either C-Vic group ($P=1.00$). Similarly, with 19 x 25 mil wires at this interval, A-Emp(B) generated a torquing moment that was not significantly different from the C-Vic(P) group ($P=1.00$) or A-Spd(B) group ($P=0.806$). With 20 x 20 mil wires at this interval, the C-Vic groups alone generated the second highest mean torquing moments, which were significantly different from all other groups ($P<0.05$). An exception to this pattern was observed with 16 x 22 mil wires, which showed numerous non-significant differences in mean torquing moments generated at 30 degrees between groups aside from the P-Dmn groups.

At 45 degrees of rotation with the smallest dimension wire tested (16 x 22 mil), again both P-Dmn groups demonstrated significantly greater mean torquing moments than all other bracket systems ($P<0.001$). The second largest torquing moment at this wire size-rotation combination was generated by the A-Spd(B) group, and this moment was significantly different from all other groups ($P<0.001$). Similarly, with 17 x 25 mil wires at 45 degrees of rotation, both P-Dmn groups exhibited significantly greater mean torquing moments than all other groups ($P<0.05$). The C-Vic groups along with the A-Spd(B) group generated the second highest torquing moments, which were not significantly different from one another ($P=1.00$). In contrast, with the largest wire sizes (19 x 25 and 20 x 20 mil) at 45 degrees, C-Vic and P-Dmn groups generated mean torquing moments that were not significantly different from one another but were significantly greater than most ASL groups. For instance, at 45 degrees of rotation with 20 x 20 mil wires, P-Dmn(B) generated significantly greater mean torquing moments than all other groups ($P<0.001$) except P-Dmn(P) ($P=1.00$), C-Vic(B) ($P=1.00$), C-Vic(P) ($P=1.00$), and A-Emp(B) ($P=0.136$).

		Bracket Group									
Wire Size (mil)	Degree	P-Dmn(B)	P-Dmn(P)	A-Emp(B)	A-Emp(P)	A-Spd(B)	A-Spd(P)	A-Vic(B)	A-Vic(P)	C-Vic(B)	C-Vic(P)
16x22	15	0.71(0.19) ^{a}	1.21(0.5) ^{bcd}	1.7(0.43) ^{e}	1.23(0.39) ^{bcd}	1.27(0.21) ^{cde}	0.97(0.12) ^{abc}	1.1(0.13) ^{abcd}	0.82(0.19) ^{ab}	1.39(0.25) ^{de}	1.15(0.14) ^{bcd}
	30	4.07(0.98) ^{d}	3.17(1.12) ^{c}	1.74(0.47) ^{ab}	0.9(0.37) ^{a}	2.12(0.6) ^{b}	2.07(0.58) ^{b}	1.44(0.13) ^{ab}	1.52(0.15) ^{ab}	1.72(0.28) ^{ab}	1.57(0.2) ^{ab}
	45	22.45(1.54) ^{e}	21.81(2.33) ^{e}	9.17(1.52) ^{b}	5.65(1.37) ^{a}	17.13(1.91) ^{d}	10.07(1.29) ^{bc}	3.48(1.4) ^{a}	3.53(1.97) ^{a}	11.67(1.4) ^{bc}	11.83(2.01) ^{c}
17x25	15	3.45(0.96) ^{c}	3.38(1.66) ^{c}	2.93(0.9) ^{bc}	2.1(0.69) ^{abc}	2.56(1.72) ^{abc}	1.19(0.6) ^{a}	1.67(0.43) ^{ab}	1.57(1.16) ^{ab}	1.39(0.21) ^{a}	1.34(0.18) ^{a}
	30	34.4(1.42) ^{e}	33.71(1.75) ^{e}	23.57(2.3) ^{c}	13.19(1.04) ^{b}	29.58(3.5) ^{d}	12.35(1.99) ^{ab}	19.81(2.71) ^{c}	9.28(3.81) ^{a}	28.93(2.44) ^{d}	27.59(2.76) ^{d}
	45	65.35(1.78) ^{e}	64.69(1.53) ^{e}	53.07(2.37) ^{c}	33.17(1.22) ^{b}	59.21(2.28) ^{d}	25.71(2.35) ^{a}	51.69(3.04) ^{c}	30.92(4.14) ^{b}	60.54(2.22) ^{d}	60.93(2.73) ^{d}
19x25	15	16.3(1.52) ^{e}	15.25(0.71) ^{e}	9.58(1.13) ^{bcd}	7.37(1.75) ^{ab}	11.22(2.9) ^{d}	8.21(1.85) ^{bc}	7.64(1.98) ^{ab}	5.13(1.76) ^{a}	11.27(0.81) ^{d}	10.41(1.81) ^{cd}
	30	49.47(2.19) ^{f}	49.83(2.09) ^{f}	38.24(1.73) ^{d}	29.33(1.83) ^{bc}	41.17(4.27) ^{de}	22.76(2.46) ^{a}	32.63(3.75) ^{c}	25.54(2.92) ^{ab}	42.63(1.77) ^{e}	40.27(2.89) ^{de}
	45	80.87(1.64) ^{e}	79.83(2.48) ^{e}	70.08(1.81) ^{cd}	51.15(1.84) ^{b}	45.38(19.48) ^{b}	34.06(4.02) ^{a}	64.74(4.28) ^{c}	53.8(3.79) ^{b}	75.77(1.79) ^{de}	72.9(3.62) ^{cde}
20x20	15	17.4(1.36) ^{d}	18.2(1.47) ^{d}	10.84(1.7) ^{bc}	9.02(1.45) ^{b}	10.73(1.67) ^{bc}	2.93(1.88) ^{a}	4.22(2.01) ^{a}	3.08(2.05) ^{a}	12.91(2.92) ^{c}	13.07(1.01) ^{c}
	30	51.73(1.15) ^{f}	51.63(1.45) ^{f}	41.78(2.46) ^{d}	30.39(2.35) ^{c}	40.88(2.22) ^{d}	13.88(2.74) ^{a}	31.25(3.93) ^{c}	23.35(4.23) ^{b}	46.2(3.81) ^{e}	46.49(2.02) ^{e}
	45	75.95(0.81) ^{d}	74.82(1.11) ^{d}	65.73(2.86) ^{cd}	46.45(3.55) ^{b}	21.46(21.69) ^{a}	16.55(3.15) ^{a}	58.89(3.69) ^{c}	43.41(5.34) ^{b}	71.61(3.31) ^{d}	71.01(2.48) ^{d}

Table 3: Mean torquing moments in Nmm (±SD) generated for each bracket system-wire combination for every 15-degree increment of rotation from 0 through 45 degrees. Non-significant differences between different bracket groups within a wire size-rotational increment at P>0.05 are denoted by shared alphabetical letters within each row

Wire Size (mil)	Degree	Significance Results
16x22	15	F(9,90) = 9.85, P<0.001
	30	F(9,90) = 25.04, P<0.001
	45	F(9,90) = 163.02, P<0.001
17x25	15	F(9,90) = 7.54, P<0.001
	30	F(9,90) = 132.17, P<0.001
	45	F(9,89) = 358.57, P<0.001
19x25	15	F(9,90) = 40.69, P<0.001
	30	F(9,90) = 119.43, P<0.001
	45	F(9,87) = 62.97, P<0.001
20x20	15	F(9,90) = 91.56, P<0.001
	30	F(9,90) = 200.93, P<0.001
	45	F(9,90) = 85.02, P<0.001

Table 4: Significance results comparing mean torquing moments between bracket systems within a wire size-rotational increment

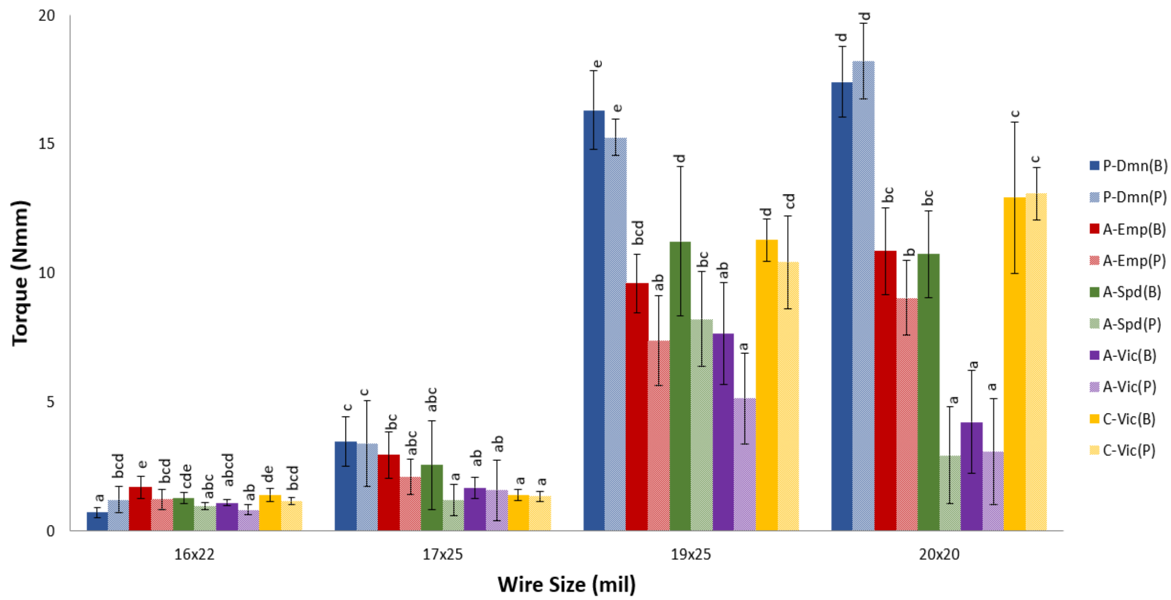


Figure 24: Mean torquing moments in Nmm (±SD) measured at 15 degrees with different bracket systems versus wire size. Error bars represent 1 SD, and letters shared within each wire size cluster represent non-significant differences between bracket systems at P>0.05

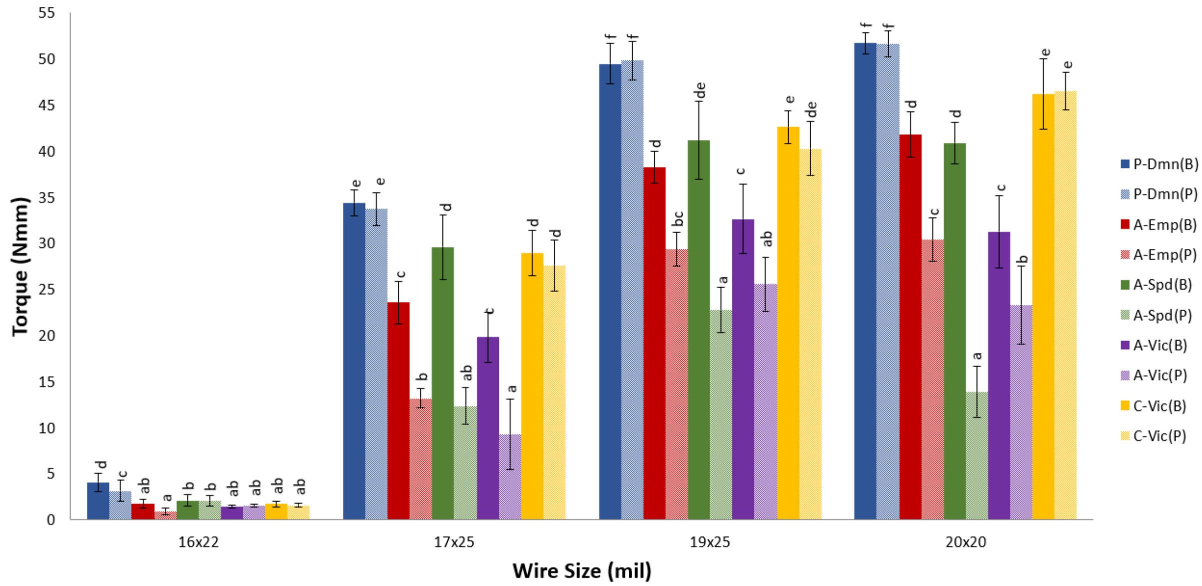


Figure 25: Mean torquing moments in Nmm (±SD) measured at 30 degrees with different bracket systems versus wire size. Error bars represent 1 SD, and letters shared within each wire size cluster represent non-significant differences between bracket systems at P>0.05

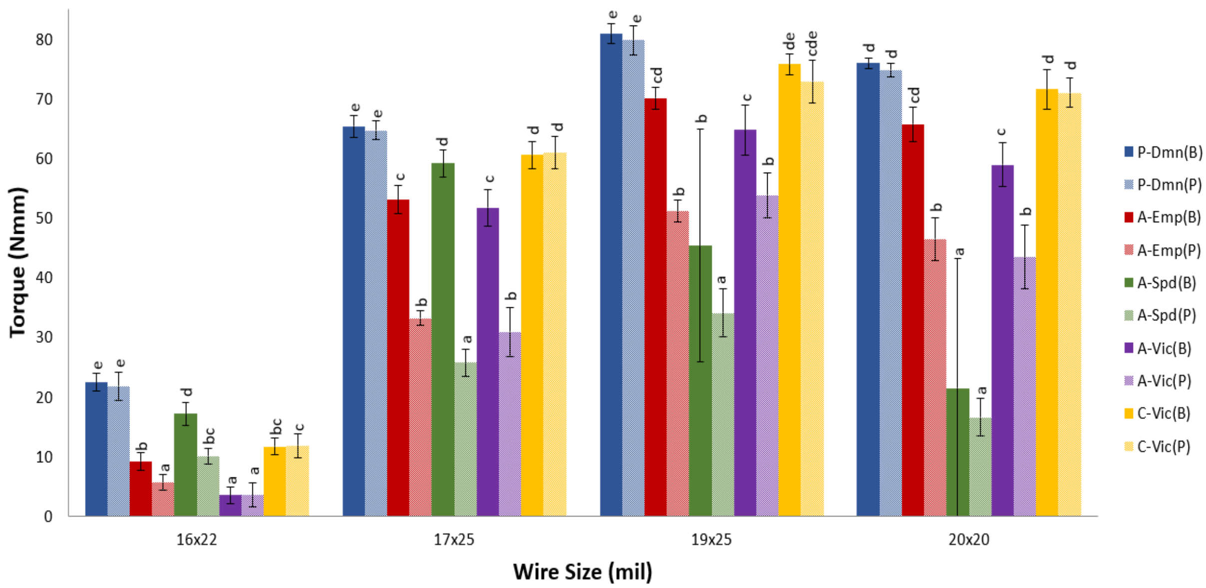


Figure 26: Mean torquing moments in Nmm (±SD) measured at 45 degrees with different bracket systems versus wire size. Error bars represent 1 SD, and letters shared within each wire size cluster represent non-significant differences between bracket systems at P>0.05

4.1.2 Comparing Wire Sizes

Comparing mean torquing moments generated by a given bracket system with different wire sizes at a given degree of rotation demonstrated a statistically significant difference between groups for all comparisons. Associated means (±SD) along with significance results are highlighted in both Table 5 and Figures 27-29. Associated P-values are found in Table 6.

More specifically, 16 x 22 mil wires generally produced significantly lower mean torquing moments than the other three wire sizes, regardless of bracket group or degree of rotation, although in some instances the mean torquing moments generated with these wires were not significantly different than those generated with 17 x 25 mil wires, especially at fewer degrees of rotation. For instance, with the A-Emp(B) group at 15 degrees of rotation, mean torquing moments generated with a 16 x 22 mil wire were significantly lower than were generated with other wire sizes ($P<0.001$) except as compared to mean torque generated with 17 x 25 mil wires at this interval ($P=0.119$). However, as rotation progressed to 30 through 45 degrees for this group, mean torquing moments generated with 16 x 22 mil wires were significantly less than all other wire sizes ($P<0.001$).

In turn, 19 x 25 and 20 x 20 mil wires both produced significantly greater mean torquing moments than either 16 x 22 or 17 x 25 mil wires, regardless of degree of rotation or bracket group ($P<0.001$). At 15 and 30 degrees of rotation, the tendency was for 19 x 25 and 20 x 20 mil wires to either produce moments that were not significantly different from one another, or for 20 x 20 mil wires to produce significantly greater mean torquing moments than 19 x 25 mil wires. However, at 45 degrees of rotation the pattern was often reversed, with 19 x 25 mil wires tending to produce significantly greater mean torquing moments than 20 x 20 mil wires. For example, considering the P-Dmn(B) group at 30 degrees of rotation, 20 x 20 mil wires generated significantly greater mean torquing moments than all other wire sizes ($P<0.001$ for all comparisons except for 19 x 25 mil wires, where $P=0.011$). In contrast, at 45 degrees of rotation in this group, 19 x 25 mil wires generated significantly greater moments than all other wire sizes ($P<0.001$).

		Wire Size (mil)			
Bracket Group	Degree	16x22	17x25	19x25	20x20
P-Dmn(B)	15	0.71(0.19)[a]	3.45(0.96)[b]	16.30(1.52)[c]	17.40(1.36)[c]
	30	4.07(0.98)[a]	34.40(1.42)[b]	49.47(2.19)[c]	51.73(1.15)[d]
	45	22.45(1.54)[a]	65.35(1.78)[b]	80.87(1.64)[d]	75.95(0.81)[c]
P-Dmn(P)	15	1.21(0.5)[a]	3.38(1.66)[b]	15.25(0.71)[c]	18.20(1.47)[d]
	30	3.17(1.12)[a]	33.71(1.75)[b]	49.83(2.09)[c]	51.63(1.45)[d]
	45	21.81(2.33)[a]	64.69(1.53)[b]	79.83(2.48)[d]	74.82(1.11)[c]
A-Emp(B)	15	1.70(0.43)[a]	2.93(0.9)[a]	9.58(1.13)[b]	10.84(1.7)[b]
	30	1.74(0.47)[a]	23.57(2.3)[b]	38.24(1.73)[c]	41.78(2.46)[d]
	45	9.17(1.52)[a]	53.07(2.37)[b]	70.08(1.81)[d]	65.73(2.86)[c]
A-Emp(P)	15	1.23(0.39)[a]	2.10(0.69)[a]	7.37(1.75)[b]	9.02(1.45)[c]
	30	0.90(0.37)[a]	13.19(1.04)[b]	29.33(1.83)[c]	30.39(2.35)[c]
	45	5.65(1.37)[a]	33.17(1.22)[b]	51.15(1.84)[d]	46.45(3.55)[c]
A-Spd(B)	15	1.27(0.21)[a]	2.56(1.72)[a]	11.22(2.9)[b]	10.73(1.67)[b]
	30	2.12(0.6)[a]	29.58(3.5)[b]	41.17(4.27)[c]	40.88(2.22)[c]
	45	17.13(1.91)[a]	59.21(2.28)[b]	45.38(19.48)[b]	21.46(21.69)[a]
A-Spd(P)	15	0.97(0.12)[a]	1.19(0.6)[a]	8.21(1.85)[c]	2.93(1.88)[b]
	30	2.07(0.58)[a]	12.35(1.99)[b]	22.76(2.46)[c]	13.88(2.74)[b]
	45	10.07(1.29)[a]	25.71(2.35)[c]	34.06(4.02)[d]	16.55(3.15)[b]
A-Vic(B)	15	1.10(0.13)[a]	1.67(0.43)[a]	7.64(1.98)[c]	4.22(2.01)[b]
	30	1.44(0.13)[a]	19.81(2.71)[b]	32.63(3.75)[c]	31.25(3.93)[c]
	45	3.48(1.4)[a]	51.69(3.04)[b]	64.74(4.28)[d]	58.89(3.69)[c]
A-Vic(P)	15	0.82(0.19)[a]	1.57(1.16)[ab]	5.13(1.76)[c]	3.08(2.05)[b]
	30	1.52(0.15)[a]	9.28(3.81)[b]	25.54(2.92)[c]	23.35(4.23)[c]
	45	3.53(1.97)[a]	30.92(4.14)[b]	53.80(3.79)[d]	43.41(5.34)[c]
C-Vic(B)	15	1.39(0.25)[a]	1.39(0.21)[a]	11.27(0.81)[b]	12.91(2.92)[b]
	30	1.72(0.28)[a]	28.93(2.44)[b]	42.63(1.77)[c]	46.20(3.81)[d]
	45	11.67(1.4)[a]	60.54(2.22)[b]	75.77(1.79)[d]	71.61(3.31)[c]
C-Vic(P)	15	1.15(0.14)[a]	1.34(0.18)[a]	10.41(1.81)[b]	13.07(1.01)[c]
	30	1.57(0.2)[a]	27.59(2.76)[b]	40.27(2.89)[c]	46.49(2.02)[d]
	45	11.83(2.01)[a]	60.93(2.73)[b]	72.90(3.62)[c]	71.01(2.48)[c]

Table 5: Mean torquing moments in Nmm (±SD) generated for each bracket-wire combination for every 15-degree increment of rotation from 0 through 45 degrees. Non-significant differences between wire sizes within a bracket group-rotational increment at P>0.05 are denoted by shared alphabetical letters within each row

Bracket Group	Degree	Significance Results
P-Dmn(B)	15	F(3,36) = 578.50, P<0.001
	30	F(3,36) = 2126.82, P<0.001
	45	F(3,36) = 3184.83, P<0.001
P-Dmn(P)	15	F(3,36) = 505.95, P<0.001
	30	F(3,36) = 1868.81, P<0.001
	45	F(3,35) = 1839.82, P<0.001
A-Emp(B)	15	F(3,36) = 165.53, P<0.001
	30	F(3,36) = 907.13, P<0.001
	45	F(3,36) = 1600.46, P<0.001
A-Emp(P)	15	F(3,36) = 102.36, P<0.001
	30	F(3,36) = 787.25, P<0.001
	45	F(3,36) = 862.59, P<0.001
A-Spd(B)	15	F(3,36) = 77.69, P<0.001
	30	F(3,36) = 377.05, P<0.001
	45	F(3,33) = 17.62, P<0.001
A-Spd(P)	15	F(3,36) = 62.09, P<0.001
	30	F(3,36) = 161.26, P<0.001
	45	F(3,36) = 132.56, P<0.001
A-Vic(B)	15	F(3,36) = 43.70, P<0.001
	30	F(3,36) = 225.66, P<0.001
	45	F(3,36) = 727.03, P<0.001
A-Vic(P)	15	F(3,36) = 16.77, P<0.001
	30	F(3,36) = 128.51, P<0.001
	45	F(3,36) = 294.87, P<0.001
C-Vic(B)	15	F(3,36) = 166.39, P<0.001
	30	F(3,36) = 687.64, P<0.001
	45	F(3,36) = 1656.22, P<0.001
C-Vic(P)	15	F(3,36) = 348.27, P<0.001
	30	F(3,36) = 787.81, P<0.001
	45	F(3,36) = 1073.40, P<0.001

Table 6: Significance results comparing mean torquing moments between wire sizes within a bracket system-rotational increment

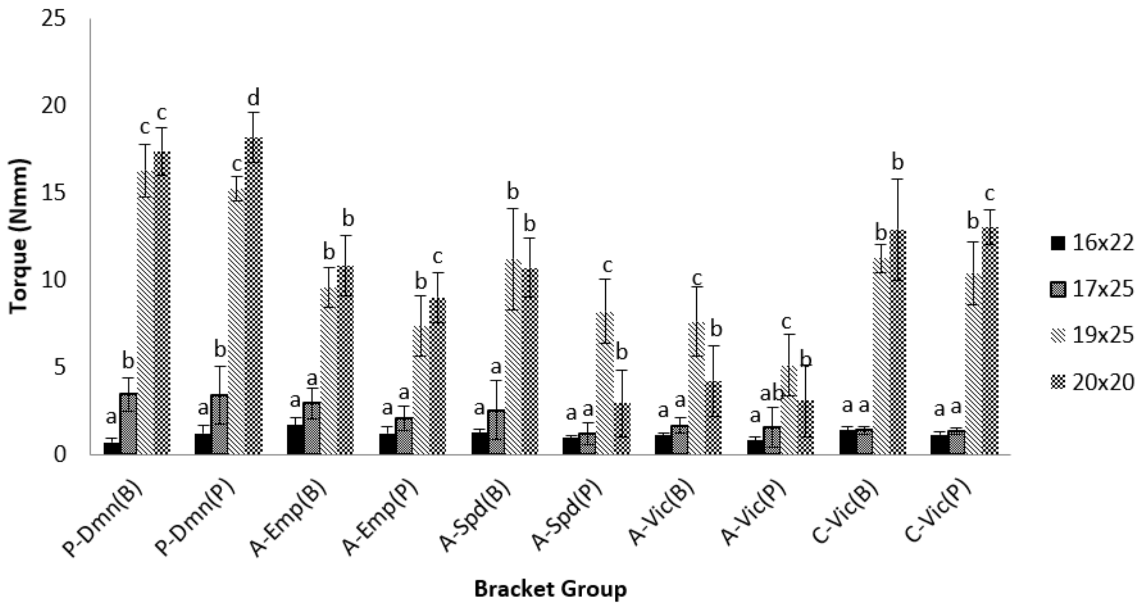


Figure 27: Mean torquing moments in Nmm (±SD) measured at 15 degrees with different wire sizes (mil) versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

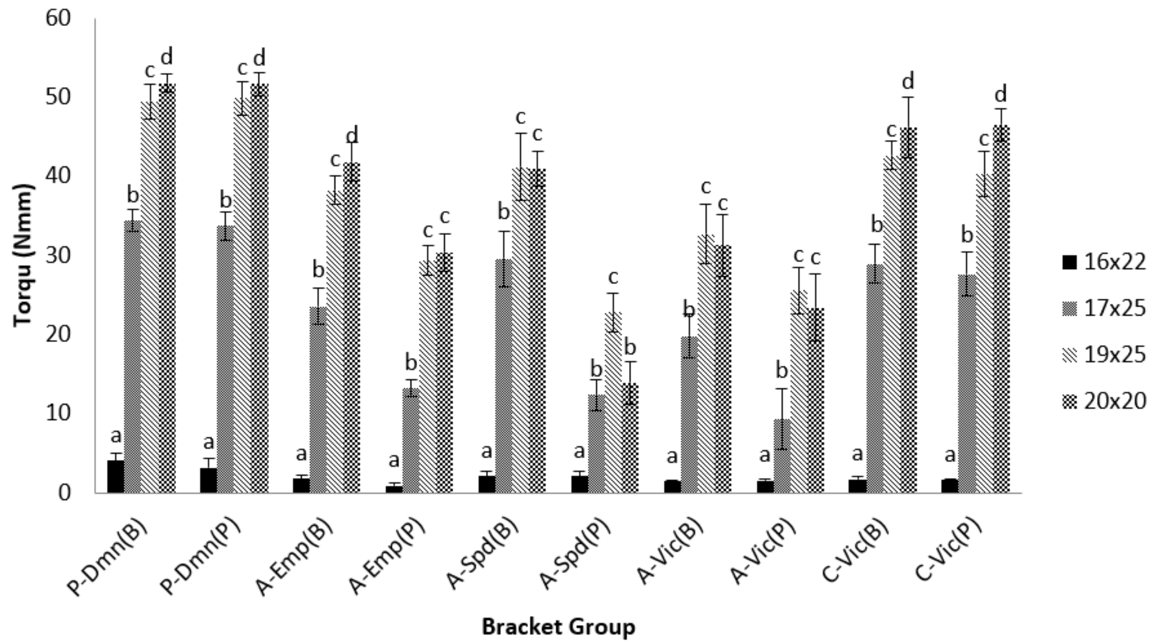


Figure 28: Mean torquing moments in Nmm (±SD) measured at 30 degrees with different wire sizes (mil) versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

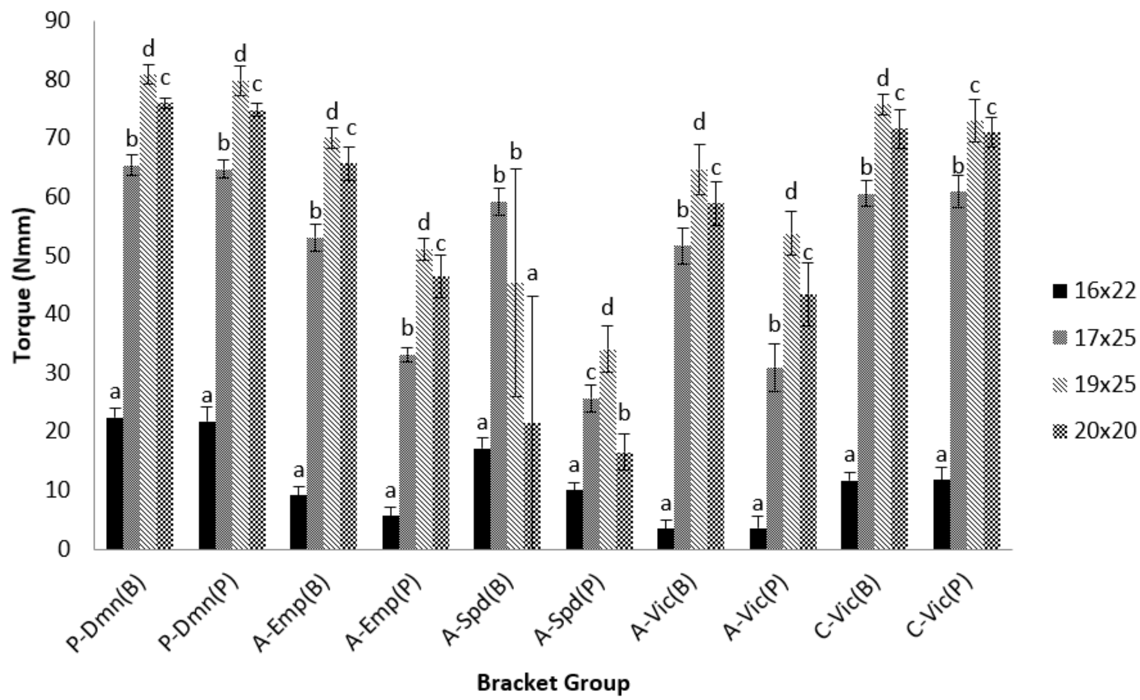


Figure 29: Mean torquing moments in Nmm (±SD) measured at 45 degrees with different wire sizes (mil) versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

4.2 Torquing Direction

Comparing each system in both buccal and palatal root torque simulations with different wire sizes revealed direction of torque simulation tended to influence ASL systems but not the PSL (P-Dmn) or conventional (C-Vic) bracket systems, especially after engagement of the wire within the bracket slot, as shown in Figures 30-34. Associated significance results are shown in Table 3, Table 4, and Figures 24-26.

Indeed, comparing mean torque generated with the P-Dmn(B) and P-Dmn(P) groups at each 15-degree increment for each wire size consistently revealed no significant difference between these two groups (P=1.00), with the exception of 15 and 30 degrees of rotation with the 16 x 22 mil wire (P=0.009 and P=0.042 between P-Dmn groups, respectively). Likewise, comparing the C-Vic(B) and C-Vic(P) groups at each 15-degree increment for each wire size revealed no significant differences between these groups at any 15-degree increment with any wire size (P=1.00). In contrast, for each ASL system, buccal and palatal root torque simulations tended to show significant differences at each 15-degree increment for each wire size, especially at higher degrees of rotation with larger wire sizes. Specifically, with these systems, mean torquing moments tended to be significantly greater in buccal root torque simulations as opposed to palatal root torque simulations. For example, considering 20 x 20 mil wires and A-Emp groups, at 45 degrees of rotation the A-Emp(B) group produced significantly greater mean torquing moments than the A-Emp(P) group (P<0.001).

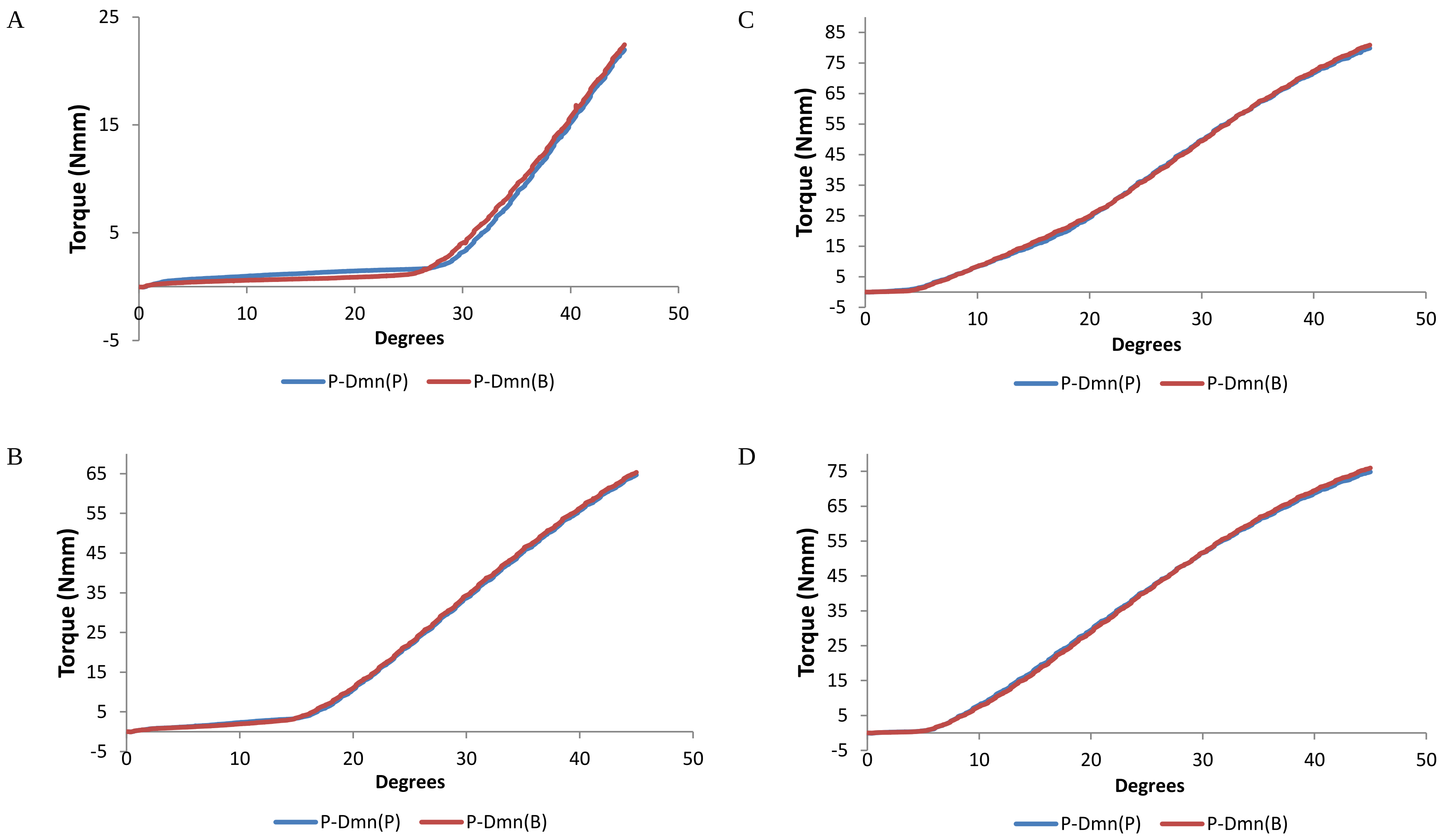


Figure 30: Mean torquing moments measured for P-Dmn brackets in both buccal (P-Dmn(B)) and palatal (P-Dmn(P)) root torque simulation directions with 16 x 22 mil SS wire (A), 17 x 25 mil SS wire (B), 19 x 25 mil SS wire (C), and 20 x 20 mil SS wire (D)

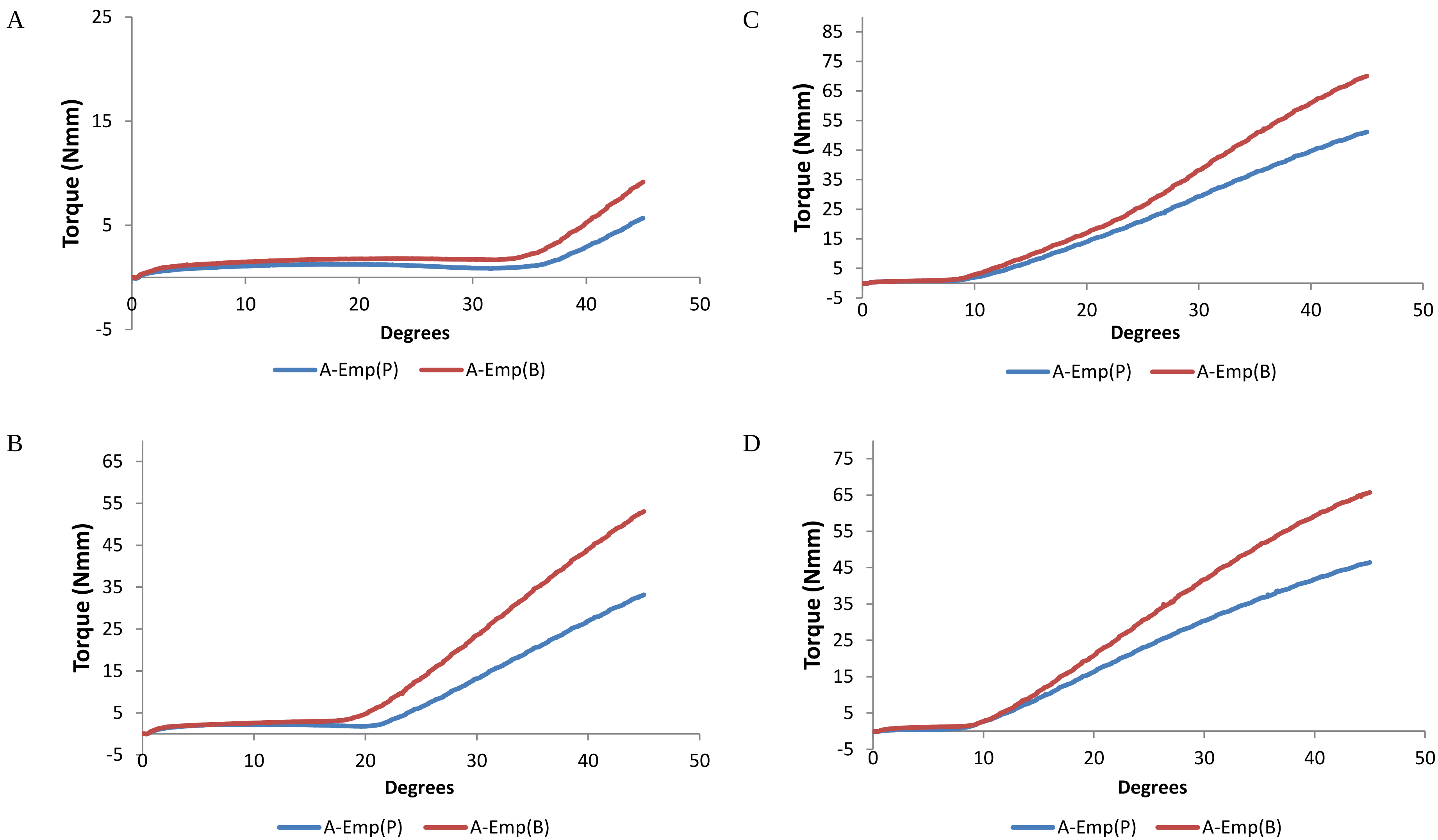


Figure 31: Mean torquing moments measured for A-Emp brackets in both buccal (A-Emp(B)) and palatal (A-Emp(P)) root torque simulation directions with 16 x 22 mil SS wire (A), 17 x 25 mil SS wire (B), 19 x 25 mil SS wire (C), and 20 x 20 mil SS wire (D)

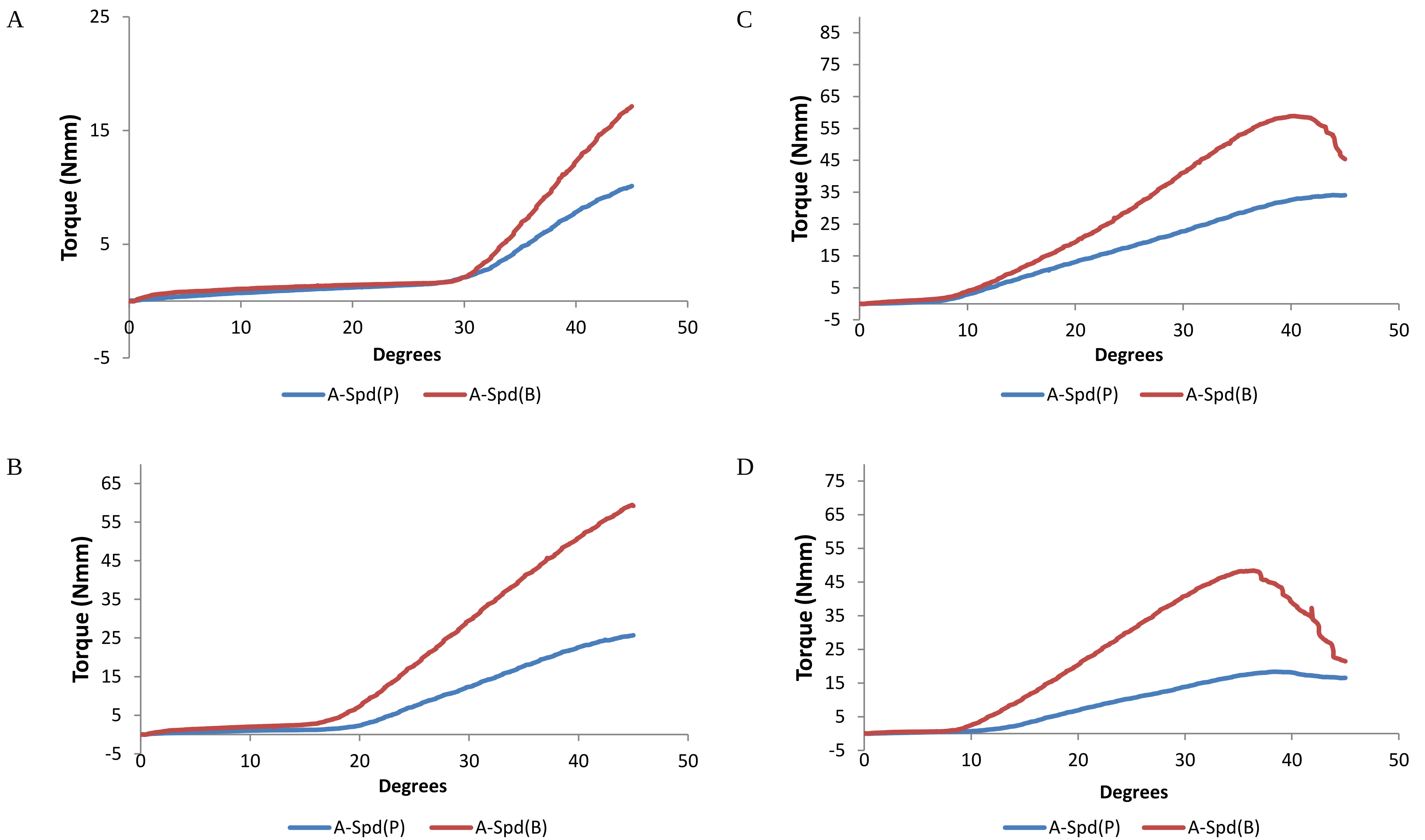


Figure 32: Mean torquing moments measured for A-Spd brackets in both buccal (A-Spd(B)) and palatal (A-Spd(P)) root torque simulation directions with 16 x 22 mil SS wire (A), 17 x 25 mil SS wire (B), 19 x 25 mil SS wire (C), and 20 x 20 mil SS wire (D)

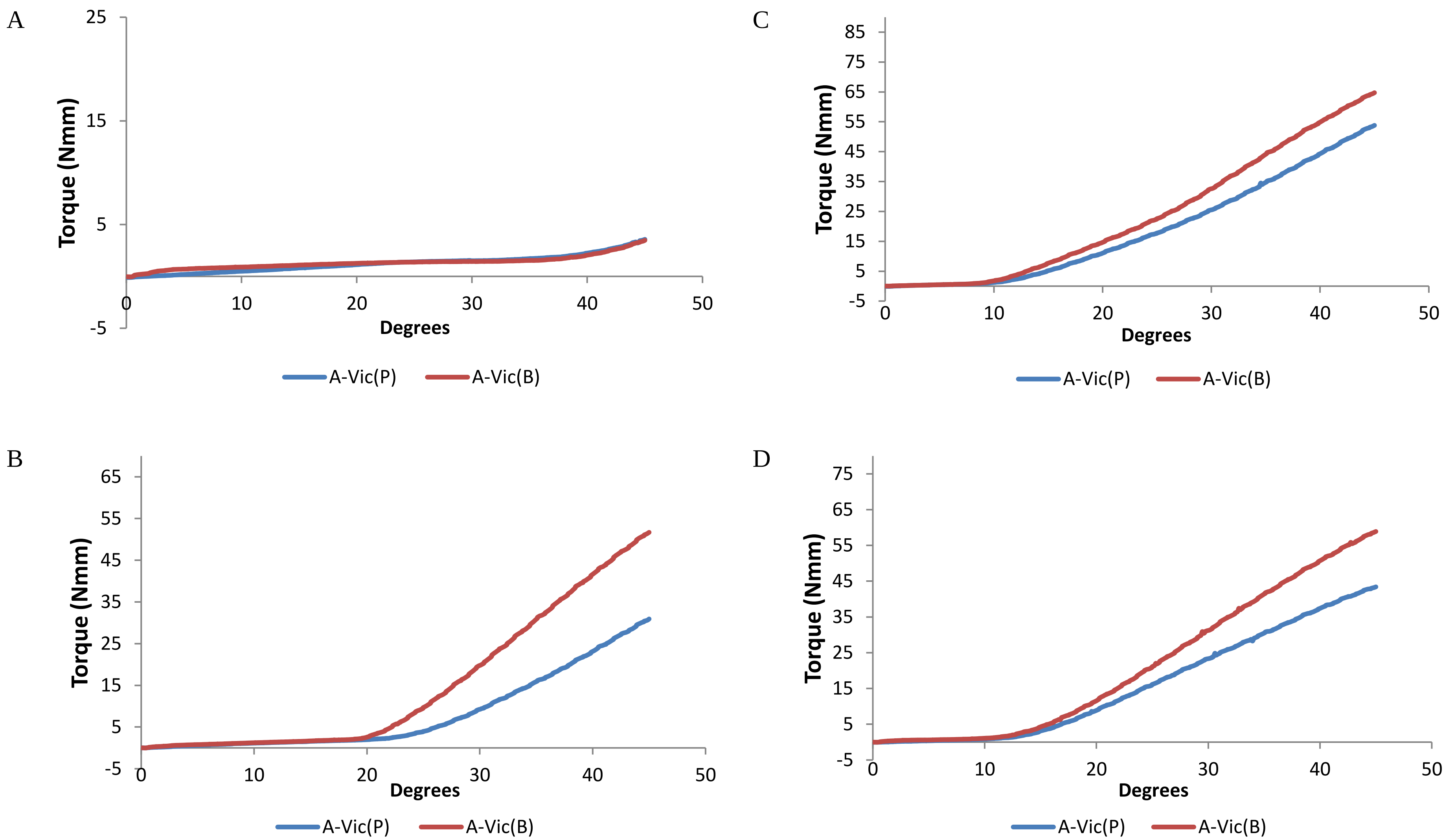


Figure 33: Mean torquing moments measured for A-Vic brackets in both buccal (A-Vic(B)) and palatal (A-Vic(P)) root torque simulation directions with 16 x 22 mil SS wire (A), 17 x 25 mil SS wire (B), 19 x 25 mil SS wire (C), and 20 x 20 mil SS wire (D)

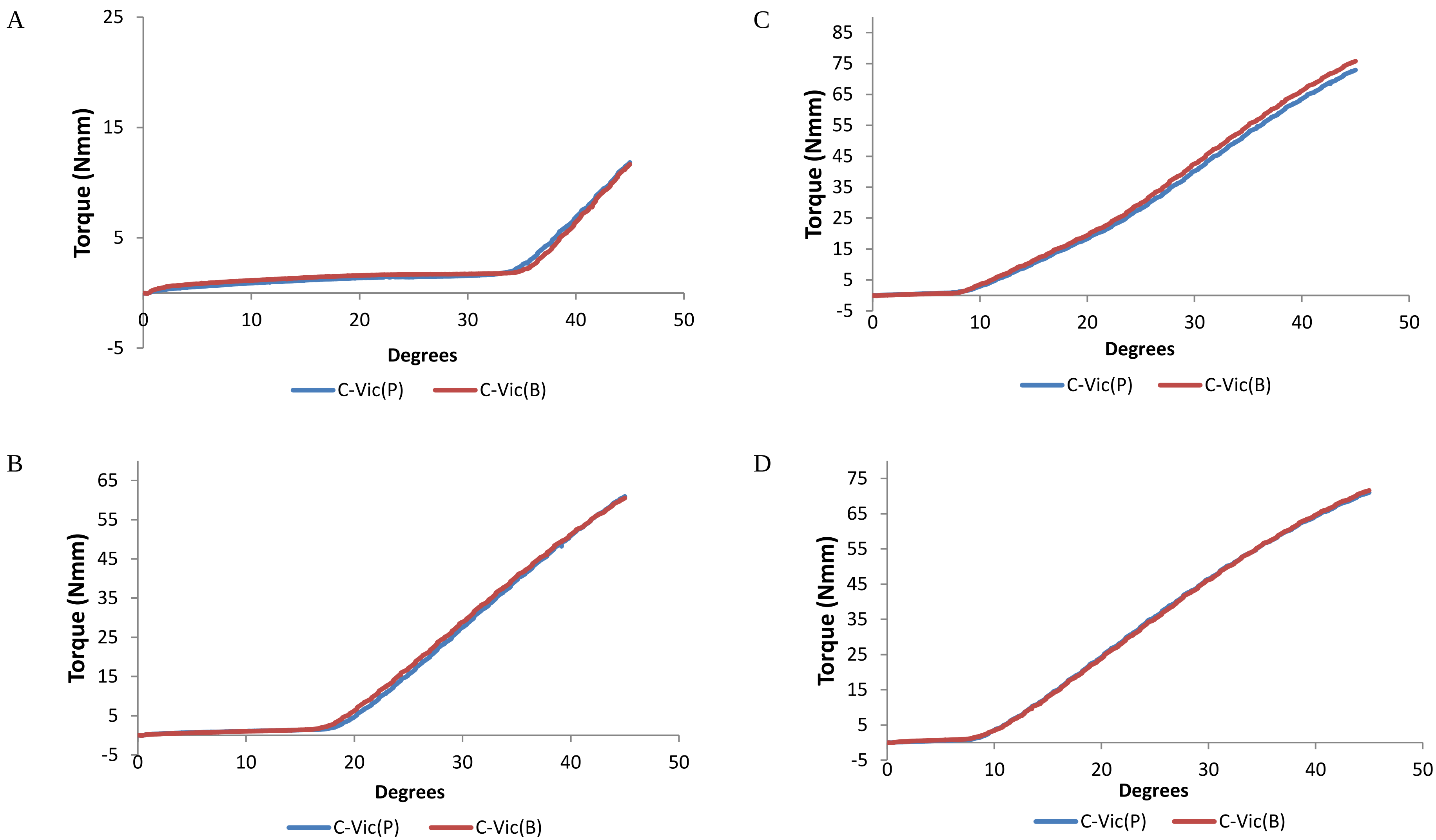


Figure 34:Mean torquing moments measured for C-Vic brackets in both buccal (C-Vic(B)) and palatal (C-Vic(P)) root torque simulation directions with 16 x 22 mil SS wire (A), 17 x 25 mil SS wire (B), 19 x 25 mil SS wire (C), and 20 x 20 mil SS wire (D)

4.3 Engagement Angles

Comparing mean engagement angles recorded for different bracket groups within a given wire size revealed a significant difference between groups for all comparisons. Associated mean engagement angles (±SD) along with significance results are highlighted in Table 7 and Figure 35. Associated P-values are found in Table 8. Comparing mean engagement angles between wire sizes within each bracket group likewise revealed a significant difference for all comparisons. Associated mean engagement angles (±SD) along with significance results are highlighted in Table 9 and Figure 36. Associated P-values are found in Table 10.

Considering 16 x 22 mil wires, several significant differences were noted, with both the smallest and largest engagement angles noted for ASL groups. Specifically, the smallest engagement angle was recorded for the A-Spd(P) group and the largest the A-Vic(B) group, which measured 23.96 and 37.52 degrees respectively. These were significantly different from one another ($P<0.001$). A similar trend was observed with 17 x 25 mil wires whereby several significant differences were noted, although with this wire size the smallest engagement angle was recorded for the P-Dmn(B) group and the largest for the A-Vic(P) group, with these significantly different from one another ($P<0.001$). As wire sizes increased to 19 x 25 and 20 x 20 mil, the P-Dmn groups consistently demonstrated significantly smaller engagement angles than all other groups ($P<0.05$). For 19 x 25 mil wires, P-Dmn(P) had the smallest recorded engagement angle, measuring 4.08 degrees while the A-Vic(P) group had the greatest, measuring 11.11 degrees. For 20 x 20 mil wires, again P-Dmn(P) had the smallest engagement angle, measuring 5.95 degrees while A-Vic(P) had the largest, measuring 13.54 degrees.

When considering comparisons between wire sizes within each bracket group, the smallest mean engagement angles were generally recorded for both 19 x 25 and 20 x 20 mil wires, while the largest engagement angles were recorded for 16 x 22 mil wires. Indeed, for all comparisons, engagement angles for 16 x 22 mil wires were significantly larger than those recorded for all other wire sizes ($P<0.05$). Engagement angles for 19 x 25 and 20 x 20 mil wires were not significantly different from one another for each bracket system ($P>0.05$), with the exception of both P-Dmn groups and the A-Vic(B) group, where engagement angles were smaller for 19 x 25 mil wires as compared to 20 x 20 mil wires, with P=0.004, P=0.013, and P=0.027 for P-Dmn(B), P-Dmn(P), and A-Vic(B) respectively.

Concerning direction of rotation and engagement angle, whether buccal or palatal root torque was simulated did not appear to have a significant bearing on engagement angle for a given bracket system, given (B) and (P) bracket groups tended to not show significant differences in engagement angle within a given wire size (Table 7). In example, considering 20 x 20 mil wires and the A-Emp groups, mean engagement angles measured 9.19 and 8.69 degrees for the A-Emp(B) and A-Emp(P) groups respectively, with these measurements determined to be not significantly different from one another (P=1.00). An exception to this finding was the A-Vic groups paired with 17 x 25 mil wires, where A-Vic(B) and A-Vic(P) groups demonstrated significantly different mean engagement angles (P=0.032).

	Bracket System									
Wire Size (mil)	P-Dmn(B)	P-Dmn(P)	A-Emp(B)	A-Emp(P)	A-Spd(B)	A-Spd(P)	A-Vic(B)	A-Vic(P)	C-Vic(B)	C-Vic(P)
16x22	27.04(1.21) ab	28.22(1.41) abc	33.35(1.42) cde	35.36(1.43) e	29.04(1.22) abcd	23.96(4.12) a	37.52(5.28) e	30.71(11.82) bcde	33.97(0.36) de	33.37(1.37) cde
17x25	13.84(0.63) a	14.24(1.83) ab	17.65(1.99) cde	20.26(0.84) ef	15.28(3.43) abc	18.18(1.90) cde	19.66(0.84) de	22.76(3.69) f	16.98(0.82) bcd	17.91(0.66) cde
19x25	4.59(0.80) a	4.08(0.58) a	8.31(1.03) bc	9.57(1.57) cd	8.04(1.75) bc	6.91(1.67) b	9.62(1.38) cd	11.11(1.80) d	7.98(0.47) bc	8.5(1.43) bc
20x20	6.03(0.73) a	5.95(0.91) a	9.19(0.72) b	8.69(0.84) b	9.06(0.83) b	9.34(3.05) b	12.96(1.73) c	13.54(2.13) c	8.76(1.10) b	8.59(0.40) b

Table 7: Mean engagement angles in degrees (±SD) for each bracket-wire combination. Non-significant differences between groups within a given wire size at P>0.05 are denoted by shared alphabetical letters within each row

Wire Size (mil)	Significance Results
16x22	F(9,81) = 11.54, P<0.001
17x25	F(9,90) = 19.72, P<0.001
19x25	F(9, 90) = 27.00, P<0.001
20x20	F(9, 90) = 28.22, P<0.001

Table 8: Significance results comparing engagement angles between bracket systems within a given wire size

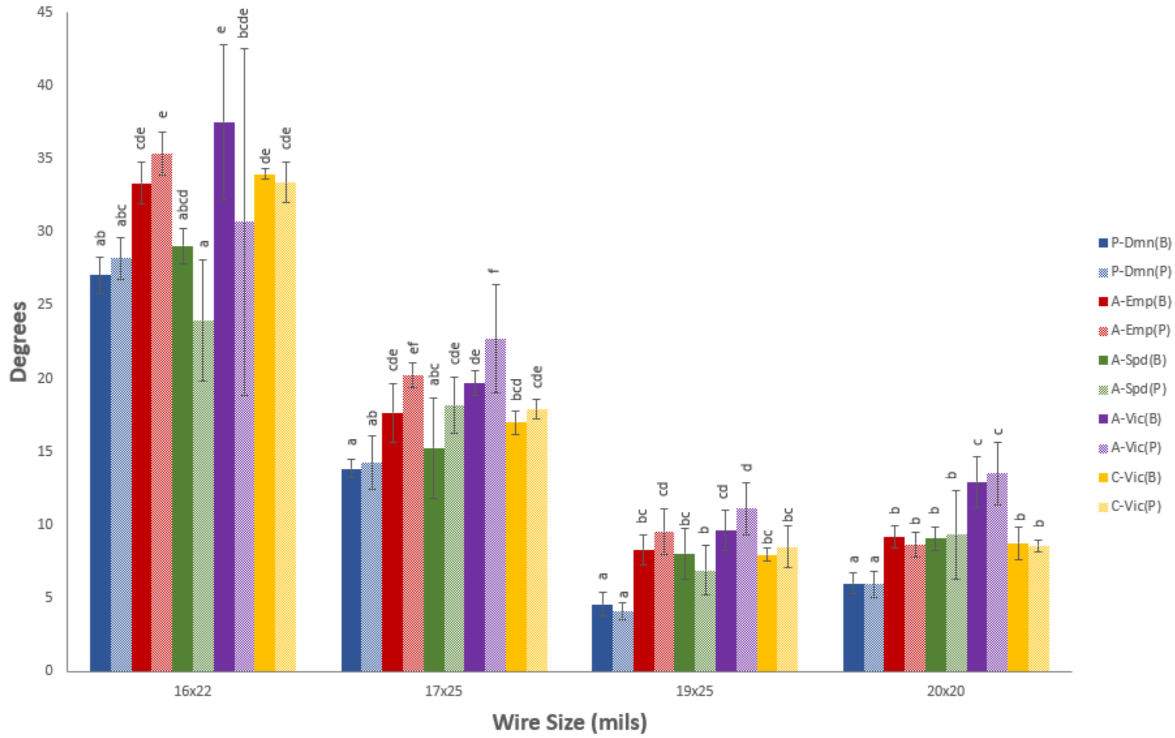


Figure 35: Mean engagement angles in degrees (±SD) with different bracket systems versus wire size. Error bars represent 1 SD, and letters shared within each wire size cluster represent non-significant differences between bracket systems at P>0.05

	Wire Size (mil)			
Bracket group	16x22	17x25	19x25	20x20
P-Dmn(B)	27.04(1.21)^{d}	13.84(0.63)^{c}	4.59(0.80)^{a}	6.03(0.73)^{b}
P-Dmn(P)	28.22(1.41)^{d}	14.24(1.83)^{c}	4.08(0.58)^{a}	5.95(0.91)^{b}
A-Emp(B)	33.35(1.42)^{c}	17.65(1.99)^{b}	8.31(1.03)^{a}	9.19(0.72)^{a}
A-Emp(P)	35.36(1.43)^{c}	20.26(0.84)^{b}	9.57(1.57)^{a}	8.69(0.84)^{a}
A-Spd(B)	29.04(1.22)^{c}	15.28(3.43)^{b}	8.04(1.75)^{a}	9.06(0.83)^{a}
A-Spd(P)	23.96(4.12)^{c}	18.18(1.90)^{b}	6.91(1.67)^{a}	9.34(3.05)^{a}
A-Vic(B)	37.52(5.28)^{d}	19.66(0.84)^{c}	9.62(1.38)^{a}	12.96(1.73)^{b}
A-Vic(P)	30.71(11.82)^{c}	22.76(3.69)^{b}	11.11(1.80)^{a}	13.54(2.13)^{a}
C-Vic(B)	33.97(0.36)^{c}	16.98(0.82)^{b}	7.98(0.47)^{a}	8.76(1.10)^{a}
C-Vic(P)	33.37(1.37)^{c}	17.91(0.66)^{b}	8.5(1.43)^{a}	8.59(0.40)^{a}

Table 9: Mean engagement angles in degrees (±SD) for each bracket-wire combination. Non-significant differences between wire sizes within a given bracket group at P>0.05 are denoted by shared alphabetical letters within each row

Bracket Group	Significance Results
P-Dmn(B)	$F(3,36) = 1400.49$, $P<0.001$
P-Dmn(P)	$F(3,36) = 745.09$, $P<0.001$
A-Emp(B)	$F(3,36) = 711.69$, $P<0.001$
A-Emp(P)	$F(3,35) = 991.21$, $P<0.001$
A-Spd(B)	$F(3,36) = 220.42$, $P<0.001$
A-Spd(P)	$F(3,36) = 76.61$, $P<0.001$
A-Vic(B)	$F(3,32) = 184.10$, $P<0.001$
A-Vic(P)	$F(3,32) =22.36$, $P<0.001$
C-Vic(B)	$F(3,36) = 2601.12$, $P<0.001$
C-Vic(P)	$F(3,36) = 1213.24$, $P<0.001$

Table 10: Significance results comparing engagement angles between wire sizes within a given bracket group

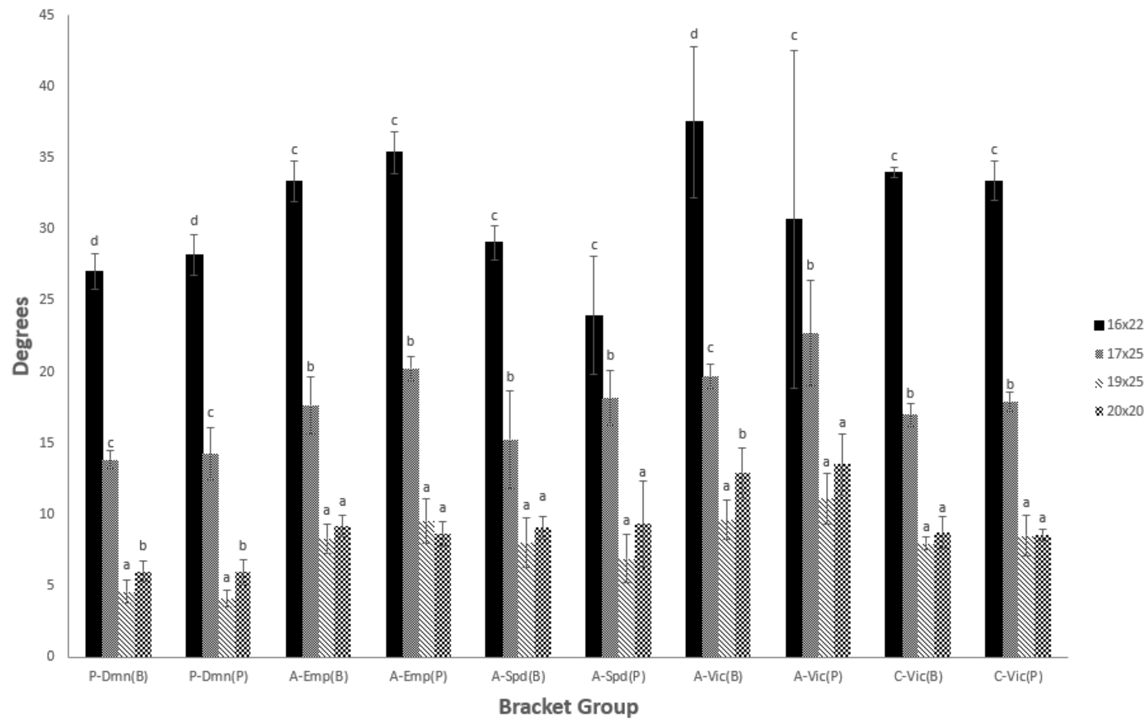


Figure 36: Mean engagement angles in degrees (±SD) with different wire sizes (mil) versus bracket system. Error bars represent 1 SD, and letters shared within each bracket group cluster represent non-significant differences between wire sizes at P>0.05

Chapter 5

5 Discussion

The primary purpose of this investigation was to assess torquing moments generated by ASL orthodontic brackets as compared to a PSL system and a conventionally ligated control in simulations of both buccal and palatal root torque with various sizes of SS wire. Thereby, this investigation aimed to build upon a previous study investigating friction generated with various bracket system-wire size combinations[76] and thus intended to explore purported benefits of ASL systems, particularly the suggestion the wire-seating mechanism of these bracket systems leads to generation of larger torquing moments for a given degree of rotation with a given wire size compared to other ligation methods, as well as reduced engagement angles for a given wire size as compared to other ligation methods. In contrast to previous studies of orthodontic torque expression, this investigation was novel in that it explored numerous dimension wires in two directions of rotation: simulations of both buccal and palatal root torque. A custom table-top torque-measuring apparatus employing 3D printing technologies was developed to undertake the investigation, and the resultant information would be expected to allow clinicians to better understand the function of the orthodontic appliances they employ, and ultimately would facilitate customization of orthodontic appliances according to the torquing needs of a given patient.

5.1 Study Methodology

Numerous methods have been developed in the past for studying torque in orthodontics. The current investigation used a novel, custom fabricated assembly to rotate the bracket system of interest around a fixed segment of wire size of interest. This methodology, although using a newly developed custom device, is comparable to methods utilized in some previous investigations, with some unique features. For example, this setup was similar to the apparatus originally developed by Badawi et al[12] and subsequently utilized in other investigations in adapted forms[21,48,58,59,81], which used a digital inclinometer to measure angle of torsion of wires, an alignment assembly to ensure alignment of bracket slot and wire, and worm-gear to rotate a wire segment. Likewise, the current setup can be considered similar to the apparatus used in completion of a previous thesis, which used an Instron E10000 to measure torquing moments generated with rotation of a fixed segment of wire.[60] Both of these previous setups similarly utilized wire spans of 15mm, as in the current investigation, which is a critical aspect for comparing procedures given the relationship between wire length and stiffness.[1]

The current methodology was different from these previous setups, however, in that the bracket was rotated while the wire of interest was held stationary, as opposed to applying torsion to the wire segment while the bracket was held stationary. However, one would expect similar findings regardless of whether bracket or wire was mobilized during testing. The rationale for this expectation is that mechanically, the same process will be occurring within the bracket slot regardless of which element of the system is mobilized. Likewise, the current methodology used visual inspection to ensure alignment of wire and bracket slot. This was consistent with the apparatus used by Young et al[60], but contrasted the apparatus developed by Badawi et al[12], which used turntables to adjust bracket position to ensure alignment. Nevertheless, the observation standard deviations of mean torquing moments generally appeared comparable to or less than those reported by those employing the latter apparatus for comparable wire materials, wire sizes, and torsional angle lends support to the effectiveness of the visual alignment approach used in this investigation.

Conversely, the current methodology using a fixed segment of wire should not be considered directly comparable to studies that have employed the OMSS system for torque testing. Specifically, given the OMSS uses a model of a complete dental arch with brackets bonded to teeth adjacent to the site of interest and a continuous archwire, one would expect torque values measured with this system to be lower and engagement angles to be greater than in the current investigation, given adjacent teeth in the model will provide increased archwire play.[44,45]

In contrast to some of the existing literature, the use of a small capacity load cell in the current investigation should be considered a distinct advantage that will have provided improved precision and accuracy to the current methodology. For example, a previous thesis employed a 10 kilonewton (kN) load cell, which was demonstrated to have significant measurement noise, and likely was limited in accuracy at low twist angles and small torque values.[60]

Likewise, investigating both directions of torsion should afford increased validity to the current study over some of the previous literature, which has often focused on a single torsional direction. Indeed, only a small selection of the previous literature has directly compared different ligation methods in both torquing directions.[38,60] Given in a clinical scenario both buccal and palatal root torque may be applied to a tooth and a continuous wire will be engaged throughout the dental arch, and considering the current investigation noted direction had a significant influence on torquing moments generated by some bracket systems, investigating both directions of torsion is imperative if one is to draw valid conclusions regarding torquing capabilities of a bracket system.

5.2 Mean Torquing Moments with Progressive Rotation (Torque-Rotation Curves)

Considering torquing moments generated with progressive rotation of a bracket around a wire size of interest, regardless of wire size, direction of rotation (simulating buccal or palatal root torque), or bracket system under investigation, mean torquing moments tended to increase with increasing twist after engagement of the wire within the bracket slot, as highlighted in the torque-rotation curves shown in Figures 20 to 23. This observation was consistent with the available literature.[12,21,22,27,38,53,54,58,59]

Moreover, when direct comparison with the available literature is considered (given comparable methodology, bracket system, wire size and material, and degree of twist), torquing moments recorded in the current investigation were generally comparable to those recorded in the available literature. In example, when considering loading curves with 19 x 25 SS wires, Major et al[21] found the Damon system available at the time generated a mean torquing moment of 9.6Nmm, 50.1Nmm, and 86.9Nmm at 15, 30, and 45 degrees respectively. In the current investigation, P-Dmn(B) generated a mean moment of 16.3Nmm, 49.47Nmm, and 80.87Nmm at 15, 30, and 45 degrees, respectively. Considering Major et al[21] reported standard deviations as large as 5.5Nmm with this system at these rotational intervals, these values can be considered comparable. Likewise, a previous thesis on the subject using similar methodology found Empower Interactive brackets generated a mean moment of 11.2Nmm, 47.7Nmm, and 80.6Nmm at 15, 30, and 42 degrees of rotation in a counter clockwise direction[60] (equivalent to buccal root torque in the current study), and the current investigation found mean torquing moments of 9.58Nmm, 38.24Nmm, and 70.08Nmm for the A-Emp(B) group at 15, 30, and 45 degrees respectively. These values are again comparable, especially when considering reported standard deviations. Comparisons to the available literature is complicated by differing methodologies, wire sizes, wire materials, and bracket systems of interest, as well as rotational intervals.

Critically, when considering torque-rotation curves, the relationship between moment and degree of rotation was approximately linear for 16 x 22 and 17 x 25 mil wires, but for 19 x 25 and 20 x 20 mil wires linearity between torquing moment and degree of rotation was lost beyond approximately 35 degrees of twist. Although direct comparison of mean moments recorded in this study to other investigations is complex due to differences in experimental setup and materials as noted above, comparison of general patterns is appropriate and reveals others have noted similar findings. For example, considering 19 x 25 mil SS wires in a 22 mil slot, Major et al[21] noted loss of linearity between torque and degree of twist beyond approximately 40 degrees. These authors suggest this phenomenon is likely due to plastic deformation of both wire and bracket, strain hardening of the wire, edge beveling, among other factors. Similar factors are likely at play in this investigation for the two largest wire sizes. In contrast, with 16 x 22 and 17 x 25 mil wires, the torques generated were likely inadequate to produce the deformations and alterations in wire and bracket characteristics observed with larger wire sizes, thus linearity throughout test cycles was generally maintained. This hypothesis is supported by the observation moments recorded for the smaller wire sizes were significantly less than recorded for the larger wire sizes, and given known alterations in material properties with differing wire dimensions, such as reduced stiffness with reduced wire diameter.[1]

Moreover, regarding the A-Spd groups with 19 x 25 and 20 x 20 mil wires, a peak torquing moment was observed, and in the case of buccal root torque simulations, a decline in torquing moments with high degrees of rotation was ultimately noted – this system demonstrated a sudden and premature loss of torque at high degrees of rotation with larger wire sizes. Critically, this pattern for this particular system was similarly observed in other studies.[12,21,58]

One possible explanation for the pattern observed with the A-Spd groups relates to the fact the clip mechanism with this system is made of NiTi, which conceivably could deflect with higher loads. Indeed, Badawi et al[12] noted load-deflection curves they observed for this system were consistent with what would be expected with a NiTi clip. Visual inspection of torque-rotation curves generated in the current study likewise reveal some curves resemble those one would expect of a NiTi clip (Figures 20 to 23), considering typical NiTi force-deflection curves.[1] Further in support of this explanation, anecdotally it was noted that under manual manipulation and visual inspection of A-Spd brackets interacting with 19 x 25 and 20 x 20 mil wires, the clip could be deformed laterally (the door was deformed outward, away from the bracket slot). As rotation was then removed, the door returned to an engaged position, indicating the deflection observed was likely elastic deformation of the NiTi clip. Such deflection could foreseeably lead to the rapid decline in torquing moments generated with these bracket-wire combinations as the wire would be allowed to disengage the slot.

Alternatively, others have reported some bracket systems plastically deform more readily than other bracket systems, resulting in increased torque play after loading[47,48], which could contribute to reduced third-order control. Plastic deformation of A-Spd brackets could likewise contribute to the observed decline in torquing moments with large dimension wires at high degrees of rotation in this investigation. If deformation is a significant factor in the loss of linearity observed in torque-rotation curves in the current study, reinforcing brackets with a stainless steel ligature would be expected to reduce associated torque loss, given the observation stainless steel ligatures reduce plastic deformation of orthodontic brackets, including SL systems, subjected to torque.[59] Ultimately, further investigation is warranted to fully elucidate the interaction between wires and bracket systems and explain these observations.

5.2.1 Comparing Bracket Systems

Comparing mean torques generated by the various bracket systems investigated highlighted that at 15 degrees of rotation, many non-significant differences between bracket systems were noted, especially when considering the two smallest wire sizes (16 x 22 and 17 x 25 mil). This observation was consistent with other investigations. For instance, Major et al[21] found that below 25 degrees of twist with a 19 x 25 mil SS wire with ASL (Speed and In-Ovation R) and PSL (DamonQ) brackets, the trend was for different systems to not show significant differences in torque values. Morina et al[50] likewise noted no significant difference in torquing moments generated between two conventional metallic twin brackets, Damon2, and Speed brackets at 20 degrees of torsion with a 19 x 25 mil SS wire.

Moreover, with the two smallest wire sizes investigated, although some significant differences were noted at the 15 degree rotational interval, the clinical significance of these differences is uncertain. Specifically, for 16 x 22 mil wires, the lowest torque measured at this interval was recorded for the P-Dmn(B) group and measured only 0.71Nmm while the largest torquing moment with this wire size was recorded for A-Emp(B) and measured 1.70Nmm. Given 5Nmm of torque is generally regarded as the threshold for orthodontic tooth movement[20,26,27], it is unlikely any bracket group would effectively provide root torque with this wire size at 15 degrees of wire twist, despite significant differences between some groups. Likewise, for 17 x 25 mil wires, the smallest torquing moment recorded at the 15 degree interval was for the A-Spd(P) group and measured 1.19Nmm while the largest was for the P-Dmn(B) group and measured 3.45Nmm, which similarly questions the clinical relevance of differences between groups with this degree of twist and 17 x 25 mil wires, given no bracket systems demonstrated a mean torquing moment above a 5Nmm threshold. Interestingly, this trend continued for 16 x 22 mil wires at 30 degrees of twist – all groups demonstrated mean torquing moments less than 5Nmm (see Table 3), questioning the clinical relevance of any significant differences between groups with this degree of twist with this wire size.

In contrast to 15 degrees of twist with 16 x 22 and 17 x 25 mil wires, and 30 degrees of twist with 16 x 22 mil wires, all other wire size-rotation interval combinations revealed at least some bracket groups generated clinically relevant moments of at least 5Nmm (see Table 3), thus the clinical relevance of significant differences between bracket systems in these comparisons can be considered. Moreover, differences in mean torquing moments between some groups for all other wire size-rotation combinations included at least some differences greater than 5Nmm, further supporting the clinical relevance of significant differences observed. As such, further examination of significant differences between groups is warranted.

Regarding general trends, it was noted that with greater degrees of twist, P-Dmn and C-Vic systems tended to produce significantly greater torquing moments than the investigated ASL systems, with some exceptions. Exceptions namely included the A-Spd and A-Emp brackets in simulations of buccal root torque at some wire size-rotation combinations. For example, with 17 x 25 mil wires at 45 degrees of twist, P-Dmn groups generated significantly larger mean torquing moments (65.35Nmm and 64.69Nmm for buccal and palatal root torque simulations, respectively) than all other systems, with the C-Vic groups and A-Spd(B) group generating the second largest mean moments (measuring 60.54Nmm, 60.93 and 59.21Nmm, for C-Vic(B), C-Vic(P), and A-Spd(B) groups respectively), which were not significantly different from one another.

This pattern was not unlike that observed in other investigations – Franco et al[77] found the Damon bracket generation available at the time (3MX) generally demonstrated significantly greater torquing moments than other tested systems; Huang et al[53] found a conventionally ligated system (Discovery) and a PSL system (Damon MX) generated significantly greater torquing moments than Speed brackets; and a previous thesis on the topic revealed similar comparisons for a similar selection of bracket systems.[60] However, others have found contrasting findings. For instance, Badawi et al[12] concluded the ASL brackets they investigated were more effective in torque control than passive systems, as did Katsikogianna et al[55]. Specifically, in the former study, the authors concluded ASL systems (In-Ovation and Speed) expressed greater torquing moments than PSL systems (Damon2 and SmartClip) at clinically useful torsion angles (up to 35 degrees of torsion), but at greater (non-clinically useful) degrees of torsion the PSL systems produced greater moments. Yet alternatively, Brauchli et al[38] investigated seven different SL bracket systems and concluded there was no significant differences between ASL and PSL systems in torque expression. Considering numerous sources of evidence together, a systematic review concluded conventionally ligated systems demonstrate greater torque expression as compared to SL systems, with ASL and PSL systems demonstrating only minor differences.[27]

The conflicting findings in the literature and in the current study regarding torque expression with different ligation methods can be at least partially explained by study methodology, particularly direction of rotation investigated. Specifically, the current investigation noted ASL systems demonstrated significantly different mean torquing moments between buccal and palatal root torque simulations, especially with greater degrees of twist, whereas PSL and conventionally ligated systems demonstrated no such directionality. Depending on whether previous studies have considered buccal or palatal root torque simulations could have considerable bearing on resultant conclusions pertaining to the influence of ligation method on torque expression as a result. Specifically, when previous studies have focused on buccal root torque simulations, ASL systems would be found to have relatively improved performance compared to other ligation methods, whereas if palatal root torque simulations were investigated, then ASL systems would demonstrate relatively poorer performance.

Likewise, aspects of the bracket design aside from ligation method likely play a significant role in torque expression, with ligation method itself likely having only a minor influence. Specifically, after investigating numerous ASL bracket systems in torque expression with the active clip in both an open and a closed position, Brauchli et al[38] concluded the spring clip contributes minimally to torque expression. Indeed, these authors concluded the spring clip exerts approximately one tenth of a clinically relevant moment, and as such concluded bracket slot dimensions are more important for torque expression than clip design. Likewise, others have noted ASL clips provide only a small force to the engaged wire, with torque expression coming almost entirely from engagement between the wire and slot walls.[21]

It follows that aspects of bracket design aside from ligation method may play a more prominent role in torque control. For example, bracket slot dimensions (depth and height) may have a significant influence on torque expression, as suggested by authors that have reported significant variation in torque generated with different replicates of a given bracket-wire combination results from structural variation in bracket slot size.[12] Likewise, the literature suggests brackets from various manufacturers may be either over or undersized versus manufacturer claims[10,19,22,25,36–39]. Variability in bracket slot dimensions could contribute to observed differences in torquing capabilities between systems if some systems demonstrate larger dimensional discrepancies versus manufacturer specifications than others. Differences between systems can also likely be partially explained by bracket width and consequently differences in free wire length – when wire segments of two different lengths are compared, stiffness can be found to increase with decreasing length[1], and stiffness of a wire will influence torque expression. A such, it can be reasoned wider brackets will produce a smaller wire span lateral to brackets and since stiffness of a wire is related to length, the wire will consequently be stiffer and generate higher torquing moments for wider brackets. In support of this theory, finite element analysis has revealed wider bracket demonstrate superior torque control as compared to narrower brackets, and increased free wire length reduces torque capabilities.[85]

Other factors which may contribute to differences in torque expression between different bracket systems include bracket material.[38,44–46,50] Although all investigated brackets in this study were metallic, finite element analysis has revealed different metallic materials show variations in deformation characteristics[49], and given deformation is known to contribute to increased torque play[47,48] differences in deformation characteristics can be expected to influence third-order control. For instance, the relatively high torquing moments the current investigation reported for P-Dmn groups could be related to the rigidity of this bracket system's design limiting deformation. This is supported by the observation Damon brackets demonstrate less deformation than other systems when subject to torquing moments.[47,48] Likewise, notching or deformation of the bracket slot walls may contribute to the findings reported in this study.[11] It follows use of a stainless steel ligature may improve torque performance of a given bracket system by reducing deformation when a third-order torque is applied.[59]

Considering these observations, the results from the current investigation, and the conflicting findings in the literature regarding the importance of ligation method on torquing moments, one may conclude many factors contribute to the overall third-order performance of a particular bracket system, including but not limited to ligation method. In other words, torque control may be more dependent upon torquing direction, the overall bracket design, and wire characteristics as opposed to intricacies of the ligation method, with rigidity and resistance to deformation of a particular bracket system likely playing a significant role. Ultimately, in this investigation, one can conclude PSL and conventionally ligated controls tended to demonstrate significantly greater torquing moments than the ASL systems examined, consistent with some of the previous literature, perhaps related to aspects of the overall bracket design, such as slot dimension, bracket rigidity, and resistance to deformation.

5.2.2 Comparing Wire Sizes

When comparing torquing moments generated by the various bracket system with different wire sizes, it was noted larger wire sizes tended to produce significantly larger torquing moments than smaller wire sizes, regardless of degree of rotation applied and bracket system investigated. This is not an exclusive observation – both Hirai et al[54] and Huang et al[53] noted torquing moments increase as SS wire sizes increase. Indeed, the latter noted the relationship between moments generated and degree of rotation are more dependent on the properties of the wire than ligation type. This observation may be at least partially related to the relationship between wire diameter and stiffness, whereby stiffness increases with increasing wire cross section.[1]

Regarding the performance of square versus rectangular wires in the current investigation, it was observed torquing moments generated by rectangular (19 x 25 mil) wires generally surpassed those generated with square (20 x 20 mil) wires at 45 degrees. These observations are corroborated by other investigations. For instance, Papageorgiou et al[33] noted rectangular wires (19 x 25 mil SS) generated statistically significantly greater moments than square wires (19 x 19 mil SS) in a 22 mil system, and Katsikogianni et al[55] found 16 x 22 mil SS wires in an 18 mil slot generated greater torquing moments than 16 x 16 mil wires. This observation is potentially related to the fact the critical distance in generating a torquing moment is the depth of the bracket-wire interaction (Figure 37), and this distance is greater for 19 x 25 mil wires (~25 mils) than 20 x 20 mil wires (~20 mils). Moreover, other authors have reported torsional stiffness of both square and rectangular wires varies considerably as a result of variations in cross-sectional geometry and material properties[15], which could have considerable bearing on torquing moments generated with different wire sizes. Yet alternatively, another investigation examining different wire sizes noted that 16 x 16 SS wires in an 18 mil slot could slip through the bracket slot (or rotate within the bracket slot), reducing torque control with these wires.[11] Given the wire sizes tested in this investigation were similarly undersized relative to the bracket slots, it is possible that at 45 degrees of rotation 20 x 20 wires begin slipping through the slot, reducing torque control with these wires at this increment.

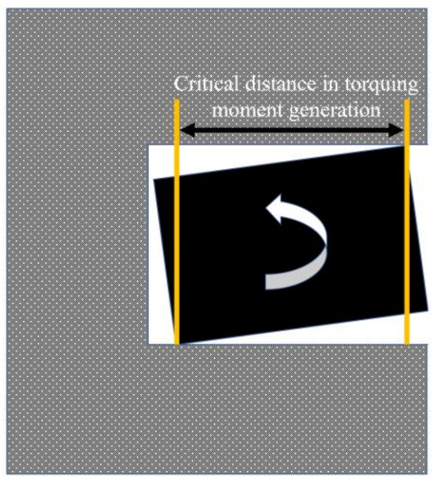


Figure 37: Critical distance in torquing moment generation, as per the formula Moment of a couple = Force x Distance. The grey patterned area represents the bracket as viewed in cross section, black represents a wire engaged in the slot, as viewed in cross-section

In contrast to the pattern observed at 45 degrees, 20 x 20 mil wires tended to generate significantly larger moments at 30 degrees of rotation as compared to 19 x 25 mil wires. This observation may again relate to various properties of the wires. For example, it is known edge bevel contributes both to engagement of the wire within the slot[14–16,23] as well as wire stiffness[23], each of which contribute to torque expression. In turn, differences in beveling between different wire sizes has been observed[23], which could contribute to differences in performance between different wire sizes at this interval. Plastic deformation of both the wire and bracket, strain hardening of the wire, among other factors have also been suggested to lead to loss of the linear relationship between torque and degree of twist with some wires[21], and it is likely different sized wires would begin to develop these alterations at different thresholds, leading to different shaped torque-rotation curves. Resultant differences in torque-rotation curves could contribute to the observation 20 x 20 mil wires tended to generate significantly larger moments at 30 degrees of rotation as compared to 19 x 25 mil wires, while the pattern was reversed at 45 degrees of twist.

Ultimately, further investigation is warranted to clarify the interactions responsible for these observed patterns, but the findings from the current investigation challenge recommendations made by some ASL bracket systems for the use of square archwires for torque control. It can be hypothesized these patterns may change when alternative wire materials are employed however, considering known differences in material properties between wire materials[1,56,57] and given torquing moments generated with TMA and NiTi wires tend to be significantly lower than those generated with SS. [22,32,53–55,58] Specifically, considering the role of material properties in wire size performance outlined above, differences in material properties associated with wire material may likewise have significant bearing on third-order performance with different wire sizes. It is possible the active clips of ASL systems may have improved performance when combined with various sizes of more flexible wire materials for this rationale.

5.3 Direction of Rotation

When comparing mean torquing moments generated by bracket systems in simulations of buccal and palatal root torque, it was observed direction of rotation tended to influence torquing moments generated by ASL systems but not the PSL or conventionally ligated systems, with palatal root torque simulations tending to produce significantly lower torquing moments than buccal root torque simulations with ASL systems. This observation corroborates those reported in other studies. Brauchli et al[38] noted significantly reduced torquing moments in a simulation of palatal root torque as opposed to buccal root torque for Speed brackets. Others have reported similar torquing moments for a conventionally ligated bracket torqued 15 degrees in both directions, although the authors did not conduct statistical analysis comparing the two directions.[31,32] Conversely, a thesis on the subject found differences in moments generated with both directions of twist regardless of bracket system investigated, although the differences were more pronounced with Speed and Empower Interactive systems as compared to DamonQ and Victory Twin brackets, consistent with the current study.[60]

The comparatively poor performance of ASL systems in simulations of palatal root torque could be a result of numerous factors. For one, authors have suggested these observations could be due to differential beveling of the wire.[38] For instance, the four edge bevels of a given wire often differ from one another and may be irregular[23], thus torque expression could differ depending on the direction of wire rotation. However, if this was a significant factor in this investigation, one would expect to note directionality for all bracket systems as all bracket systems were tested with wires supplied by the same manufacturer and of the same lot number, and this was not observed.

Alternatively, these observations could be partially explained by the design of the bracket slot – to allow the clip to seat the wire in ASL systems, the gingival portion of the slot depth is shallower than the incisal portion. As a wire twists to generate palatal root torque, it could conceivably deform the door outward and subsequently slip past this shallow gingival portion of the slot (Figure 38A), thereby limiting the torque generated with these systems, given the critical distance in torquing moment generation is the depth of the wire/slot interaction. In contrast, for expression of buccal root torque, this phenomenon would not be at play as the wire rotates in the opposite direction and thereby will not slip past this shallow slot depth (Figure 38B). This theory is supported by the observation minimum slot depths for active systems tested in this investigation all measure less than the depth of the smallest wire assessed. Specifically, the minimum slot depths for the active systems investigated in this study were previously reported to measure 18.9 mils, 15.3 mils, and 14.0 mils for Victory SL, Speed, and Empower Interactive brackets, respectively.[75] These measures are all smaller than the smallest wire depth investigated (20 mils). Further investigation is warranted to examine this possible bracket-wire interface, however.

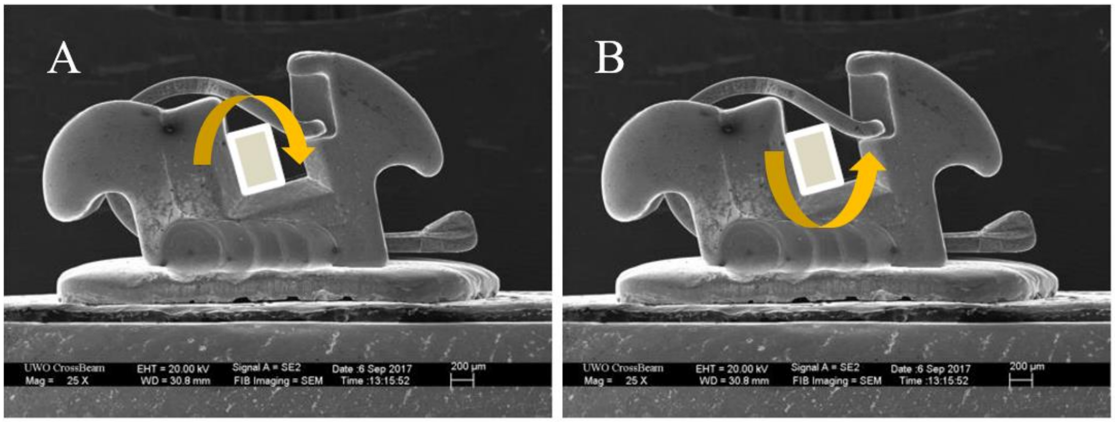


Figure 38: Scanning electron microscopy view of a self-ligating bracket, demonstrating the discrepancy in slot depth between incisal and gingival aspects to allow for seating of the active clip. (A) demonstrates the wire twist required to generate palatal root torque, (B) demonstrates the wire twist required to generate buccal root torque. Image used with permission from previous thesis completed by Greene et al[75]©

5.4 Engagement Angles

Considering engagement angles with different wire sizes within bracket groups, significantly larger mean engagement angles were noted with the smallest wire size (16 x 22mil), while the smallest mean engagement angles were noted for the largest two wire sizes (19 x 25 and 20 x 20 mil), with the latter tending to not demonstrate a significant difference between one another (Table 9 and Figure 36). This observation was consistent with the available literature demonstrating larger wire sizes are associated with reduced archwire play.[52] This finding is rational given larger wire sizes will fill the bracket slot more completely than smaller wire sizes.

Regarding direction of rotation and engagement angle, it was noted there was generally no significant difference in engagement angle when comparing buccal and palatal root torque simulations of a given bracket system (Table 7 and Figure 35). The only exception to this finding was the A-Vic groups and 17 x 25 mil wires, where A-Vic(B) and A-Vic(P) did demonstrate significantly different engagement angles. This finding could again relate to intricacies of bracket design such as minimum slot depth and clip design, as with directionality in mean torquing moments generated with ASL systems.

When comparing engagement angles between different bracket groups (Table 7 and Figure 35), with 16 x 22 mil wires no clear pattern emerged in regard to which ligation method was associated with the smallest engagement angles. Specifically, both the largest and smallest engagement angles with this wire size were recorded for ASL systems: A-Spd(P) produced the smallest engagement angle, measuring 23.96 degrees, and A-Vic(B) produced the largest, measuring 37.52 degrees, which were significantly different from one another.

As wire sizes increased, the P-Dmn groups began to demonstrate significantly smaller engagement angles compared to the other bracket groups. Specifically, for the 19 x 25 mil wires, both P-Dmn groups demonstrated engagement angles less than 5 degrees, while with 20 x 20 mil wire these two groups demonstrated engagement angles of approximately 6 degrees, with these angles significantly smaller than those recorded for all other bracket systems (Table 7 and Figure 35). This pattern corresponded to findings regarding mean torquing moments, where P-Dmn groups not only tended to demonstrate the smallest engagement angles but also the greatest torquing moments. This suggests earlier engagement may facilitate generation of larger torquing moments for a given degree of twist.

Notably, engagement angles recorded for the P-Dmn groups were smaller than theoretical estimations of engagement angles. For example, for 19 x 25 mil wires in a 22 mil slot, theoretical estimations of engagement angles are generally suggested to measure 7-9 degrees.[1,13] The measured engagement angles of less than 5 degrees with P-Dmn groups and this wire size is considerably lower than this theoretical estimation. Importantly however, all other tested bracket systems demonstrated engagement angles within or greater than this theoretical range (Table 7), consistent with the previous literature. Indeed, a systematic review of the literature noted mean engagement angles generally measure greater than theoretical engagement angles, with ASL systems on average demonstrating engagement angles of approximately 7.5 degrees and passive systems on average showing an engagement angle of approximately 14 degrees with a 19 x 25 mil wire in a 22 mil slot.[20] Considering specific comparisons between different ligation methods, Dalstra et al[24] examined a selection of 32 commercially available systems and found actual play was greater than theoretical values, especially for PSL systems. Likewise, Major et al[21] found all tested bracket systems (Speed, DamonQ, and In-Ovation R) demonstrated actual torque play larger than theoretical play.

These patterns are generally thought to result from the fact bracket slots are generally oversized versus manufacturer specifications, wires are often undersized relative to manufacturer specifications, as well as due to wire edge beveling, among other factors.[11,14,42,15,18,19,21–24,41] Likewise, the fact PSL systems do not actively press the wire into the bracket slot has been sited as a rationale for large amounts of play with these systems.[24] Conversely, some have noted ASL systems have deviation angles less than theory suggests, or less play, potentially due to the active self-ligating mechanism.[12,53]

The reduced engagement angles with the passive system observed in this study could be the result of numerous factors. For one, previous investigations have noted achieving an initial true zero position with PSL systems may be more challenging than with ASL systems.[21] Due to the wire seating mechanism of ASL systems there is high repeatability of a zero initial position as only one initial zero position should exist. With PSL systems there is considerable freedom of the wire in the bracket slot, thus a zero initial reading can occur in many positions, even if a true zero position is not achieved (Figure 39). Given that in the present investigation initial alignment of wire and slot was achieved by visual inspection, it could be argued passive systems may have had some misalignment before initiating a test cycle. Indeed, previous literature has noted actual torsional play may be substantially less than that based on theory in the presence of misalignment between wire and bracket slot.[82]

However, support for adequate alignment between wire and slot comes from the finding that tested bracket systems tended not to show a statistically significant difference when comparing engagement angles in buccal and palatal root torque simulations. Likewise, standard deviations of mean torquing moments observed for P-Dmn groups for a given wire size-rotational interval measured within the realm of other systems (Table 3), supporting consistent and accurate alignment between bracket slot and wire segment throughout the investigation – if adequate alignment was challenging one would expect to observe relatively large standard deviations for the passive system. Similarly, standard deviations of mean torquing moments within the P-Dmn groups generally remained consistent regardless of wire size (Table 5), whereas if alignment was difficult to achieve one would expect progressively larger standard deviations with smaller wire sizes as compared to larger wire sizes, further supporting the validity of the methodology.

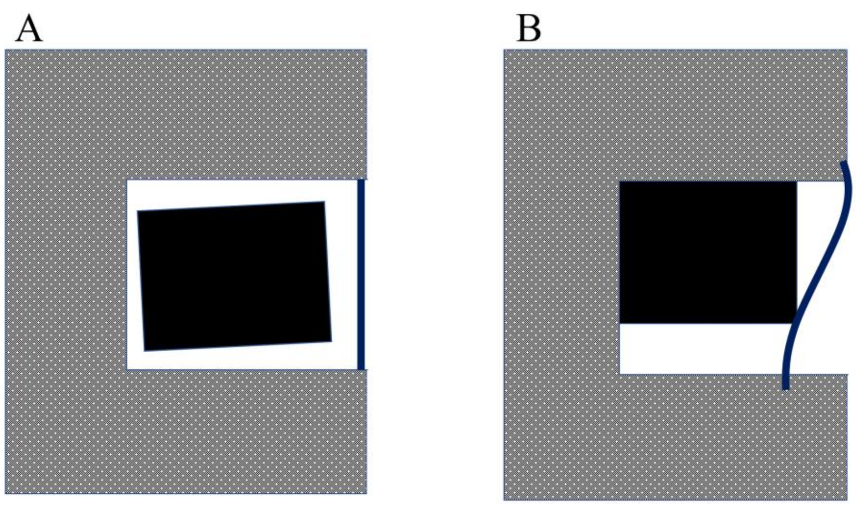


Figure 39: PSL and ASL start position diagrams. (A) demonstrates that a PSL system can have some degree of wire twist while appearing in a zero initial position, while (B) demonstrates ASL systems have a more repeatable initial starting zero position. The grey patterned area represents the bracket, black represents a wire engaged in the slot, as viewed in cross-section

Alternatively, another factor that may contribute to the comparatively small engagement angles seen with the PSL system in this study relates to slot depth. Specifically, all systems used in this investigation have a manufacturer reported slot depth of 28 mils aside from the PSL system, which has a reported slot depth of 27.5 mils. One study investigating the role of slot depth on third order control found shallower brackets may allow earlier engagement than deeper brackets, possibly due to engagement of the bracket door as opposed to the slot walls.[86] Indeed Romanyk et al[86] found that for 19 x 25 mil wires active systems engage sooner than passive systems, but for 17 x 25 mil wires passive systems with reduced slot depth engage sooner than active systems. This latter observation was consistent with the observations in the current study. Slot size may also play a related part - the observation that actual play between wire and slot is often greater than theoretical expectations is partially attributed to variations in bracket slot dimensions[11,18,22,24], and given many have reported actual slot dimensions vary from manufacturer specifications[18,24,37,41,42], one could hypothesize variability in dimensional accuracy between different bracket systems could contribute to differences in engagement angle between systems.

Finally, there are many methodologies in the literature by which engagement angles have been determined in the past. For example, some authors have used visual assessment of graphs[12,60], some have used linear regression[21], others have used formulas considering bracket slot and wire dimensions[19], among other techniques.[24] The alternative method for determining engagement angles in the present investigation complicate direct comparison of the results to the previous literature, but can perhaps be considered more valid and objective than other approaches.

5.5 Clinical Recommendations

As previously noted, clinically relevant torquing moments for biologically acceptable tooth movement are cited to range from 5 through 20 Nmm.[20,26,27] As such, it would be prudent to determine the minimum and maximum degrees of twist a clinician should incorporate into different sized wires to generate torquing moments within this biologically acceptable range.

Regarding generation of minimum moments, with 16 x 22 mil wires, twists at least as great as 31 degrees were required to generate 5Nmm of torque (P-Dmn(B)), with some bracket systems never generating this minimum threshold with this wire size (A-Vic groups), as shown in Table 11. As wire sizes increased, the trend was for progressively fewer degrees of twist to reach clinically significant torquing moments. Notably however, regardless of wire size, the degree of twist required to generate a clinically significant moment was often noted to be greater than commonly incorporated bracket prescriptions (+7 for Andrews, +12 for Roth, +17 for MBT), questioning the clinical relevance of different bracket prescriptions with the wire sizes investigated in this study. This is not a unique observation. Others have noted that torque play alone may be enough to negate incorporated torque prescriptions[19,38,50], and this is supported by clinical investigations that have reported bracket prescriptions have no effect on subjective esthetic judgements of post treatment outcomes.[63] Similarly, a recent study found bracket prescription had no clinical influence on treatment outcomes.[64]

In turn, with further rotation and large wire sizes, moments much larger than the biological threshold were generated for most bracket-wire combinations in this study. For example, the largest torquing moment recorded in the current investigation was 80.87Nmm at 45-degrees of rotation with 19 x 25 mil SS wires in the P-Dmn(B) group. Indeed, most bracket-wire combinations surpassed clinically relevant torquing moments beyond a certain degree of rotation, as shown in Table 3. Similarly, with large degrees of twist, especially with larger dimension wires, practical limitations come into play – with large degrees of twist and large dimension wires, it becomes practically impossible to engage the wire in the slot and/or close the bracket door. As a result, some simulations in the current investigation are not necessarily representative of a clinical situation.

Given both the biological and practical limitations discussed above, the degrees of twist required to generate biologically acceptable moments of 5 to 20Nmm are summarized in Table 11 and can be considered an approximate guide for the degree of twist a clinician should incorporate into a given wire size to generate a clinically relevant torquing moment in either direction of torque. Critically, these are likely underestimations however, given the experimental design used in this investigation. Specifically, in this study, the wire was firmly clamped on either side of the bracket of interest, whereas in a clinical scenario the wire is held by other orthodontic brackets on either side of a bracket of interest. This would add additional play into the system, increasing the degree of twist required to generate a clinically relevant moment in a clinical scenario. Likewise, with bracket systems demonstrating significant difference in torque expression depending on direction of twist, one would expect a clinical scenario to involve moments measuring between the two directions due to the fact the wire is tied into multiple teeth, which will dissipate torquing moments into an average value. Nonetheless, the clinician can consider these recommendations as a rough guide.

Finally, the practicing clinician should note the relatively narrow ranges of twist required to generate a clinically relevant moment (Table 12). Again these are likely underestimations given the methodology used in the current study, but nonetheless this information indicates significant care should be taken when adding torque to archwires to avoid adding an excessive degree of twist and as such generating a torquing moment in excess of what is considered clinically appropriate, which could potentially lead to injury.[28,29]

		Bracket System																			
		P-Dmn(B)		P-Dmn(P)		A-Emp(B)		A-Emp(P)		A-Spd(B)		A-Spd(P)		A-Vic(B)		A-Vic(P)		C-Vic(B)		C-Vic(P)	
Torquing Moment Generated (Nmm)		5	20	5	20	5	20	5	20	5	20	5	20	5	20	5	20	5	20	5	20
Wire Size (mil)	16x22	31.05	43.2	31.84	43.54	39.83	N/A	43.88	N/A	33.41	N/A	35.78	N/A	N/A	N/A	N/A	N/A	38.59	N/A	38.03	N/A
	17x25	16.42	23.96	16.87	24.19	20.25	28.35	23.85	34.88	18.45	25.87	22.95	37.35	22.16	30.15	26.21	38.03	19.35	26.21	20.14	27
	19x25	7.76	17.21	7.65	18	11.59	21.94	13.05	24.19	10.91	20.25	12.04	27.22	13.05	23.51	14.85	26.55	11.14	20.25	11.48	20.92
	20x20	8.55	16.2	8.44	15.86	11.59	19.57	12.04	22.39	11.59	19.8	17.55	N/A	15.64	24.3	16.87	27.67	11.02	18.22	10.91	18.22

Table 11: Degree of rotation required to generate mean torquing moments of both 5 and 20 Nmm for each bracket-wire combination examined. N/A indicates the threshold torque value was not reached by a given bracket system

		Bracket System									
		P-Dmn(B)	P-Dmn(P)	A-Emp(B)	A-Emp(P)	A-Spd(B)	A-Spd(P)	A-Vic(B)	A-Vic(P)	C-Vic(B)	C-Vic(P)
Wire size (mil)	16x22	12.15	11.7	N/A	N/A	N/A	N/A	N/A	N/A	N/A	N/A
	17x25	7.54	7.32	8.1	11.03	7.42	14.4	7.99	11.82	6.86	6.86
	19x25	9.45	10.35	10.35	11.14	9.34	15.18	10.46	11.7	9.11	9.44
	20x20	7.65	7.42	7.98	10.35	8.21	N/A	8.66	10.8	7.2	7.31

Table 12: Range of degrees of rotation required to generate a clinically relevant mean torquing moment for each bracket-wire combination examined, as determined by subtracting the degrees required to generate 20 Nmm from degrees required to generate 5Nmm of torque. N/A indicates at least one threshold torque value was not reached by a given bracket system so a range of twist could not be established

5.6 Strengths and Limitations of the Study Design

The current study demonstrated numerous strengths, as well as some weaknesses. These strengths and limitations should be reviewed and carefully considered in interpretation of the resultant data.

One of the primary assets of the current study relates to the development of a novel torque-testing apparatus – the portable, 3D printed table-top design was accessible and affordable, which could facilitate numerous centers to utilize the same design repeatably. Repeatability of research is paramount and is facilitated with this novel assembly. Moreover, the use of a small capacity load cell will have reduced measurement noise that has been observed with previous investigations[60], improving accuracy and precision of the resultant data. Likewise, regarding the methodology of the current investigation, use of linearity equations to determine engagement angles also presents an advantage over some of the previous research, whereby engagement angles have been determined by reading off of graphs.[12,60]

The scope of the current investigation also presents a distinct advantage. This study examined all three commonly used ligation methods in clinical practice today, including multiple different commonly available ASL systems; considered four different wire sizes; and examined simulations of both palatal and buccal root torque. This extended scope allowed diverse comparisons and extensive examination of factors influencing torque expression, beyond what much of the previous literature has considered. Importantly, the comparisons among variables investigated in the current investigation can be considered valid, given the same methodology was used throughout the study, whereas comparing results between different studies could be problematic due to methodological differences. Importantly, the consistent observation in this study that direction of torque simulation influences torque expression for ASL systems suggests future studies investigating torque expression should examine both directions of twist, which to date has rarely been considered. Moreover, the directionality in torque expression highlighted by the current study suggests conclusions drawn in previous investigations should be reviewed with caution.

Regarding limitations in the current methodology, one main weakness relates to the feasibility of applying the findings from an *in vitro* investigation to an *in vivo* clinical situation. Specifically, to ensure repeatability of the experiment, the wire of interest was held on either side of the test bracket with rigid clamps. This is not comparable to an *in vivo* clinical situation, in which case the wire is held by other brackets on either side of the bracket of interest. As a result, one could expect torque generation with a given degree of rotation to be lower in a clinical situation than observed in this study. For those systems demonstrating directionality in torque expression, a clinical scenario will likely produce moments approximately averaged between the two directions for a similar rationale. Likewise, the current investigation was completed in air at room temperature, whereas in clinical applications torque will be applied to the dentition at mouth temperature in a moist environment. It has been demonstrated that moisture and heat in a simulated oral environment have a pronounced effect on force decay and permanent deformation of elastomeric ligatures[72], and therefore conventionally ligated systems may perform differently clinically. Likewise, NiTi material properties change with temperature[1], thus the door of the Speed brackets may behave differently in an *in vivo* situation. Similarly, in a clinical situation, brackets are rarely fresh when torque is applied – a recent systematic review found comprehensive orthodontic treatment of adolescents requires 30.1 months.[87] Many brackets will remain in position throughout an entire treatment period, and given brackets are known to deform with use[47,48,59], it is likely torque expression would be significantly reduced in a clinical application as compared to the results of this study due to general use. Fatigue of clips and doors in SL systems may also be at play throughout such an extended treatment period.

Additionally, as previously noted, slight misalignments between the bracket slot and wire segment of interest may have been present in some bracket-wire combinations, representing yet another potential limitation of the study design. This possibility was a result of the manner in which the bracket was aligned with the wire for testing in this investigation - wire and bracket were aligned for testing by visual inspection, which could have some possibility for error, leading to misalignments. Indeed authors have reported other couples in the system can influence torque expression.[81,82] However, these effects may be limited to rotation within torsional play. Specifically, the literature suggests once torsional play is eliminated the effect of other couples on torque is minimal[82], so comparisons of torque expression beyond engagement may be unaffected.

Moreover, even if torque expression beyond engagement of the wire within the slot is affected, this effect is likely not clinically relevant. Specifically, Romanyk et al[81] concluded that although second-order misalignment produced statistically significant differences in third-order torque expression, the magnitude of the differences was likely not clinically significant, given the range of torquing moments required for biological tooth movement, variations in biological tissues *in vivo*, and appliance tolerances. However, these misalignments could theoretically have bearing on engagement angle comparisons in this investigation, especially for passive systems. Nonetheless, this is not expected to be a significant limitation in the current investigation given observed consistency in standard deviations of mean torquing moments across most bracket groups, as well as within bracket groups across different wire sizes. Likewise, consistency in engagement angles in both directions of twist and lack of directionality in torque generation in the PSL and conventional system supports good bracket-slot alignment.

5.7 Future Directions

The possibilities for future developments and research in the field of orthodontic torque expression is plentiful. Perhaps most prudent would be further development of the custom torque-testing apparatus to allow use in a micro-CT scanner, which will allow assessment of the wire-bracket interaction during torque application. Such an investigation would provide improved insight into why certain patterns were observed in this study, such as the directionality of torque expression with SL bracket systems and apparently reduced engagement angles with passive systems. Likewise, such an application would help clarify the cause of sudden, premature torque loss seen with A-Spd groups given many authors have speculated the cause is deformation or opening of the clip with this system[12,21,58], but a study specifically investigating the cause of this pattern appears to be absent from the available literature.

Testing wires of various lengths and materials are likewise valuable future investigations, particularly from a clinical standpoint. A study examining various wire spans would allow assessment of changes in torque generation in the clinically relevant scenario of variable wire spans, given the relationship between wire length and stiffness. Likewise, testing alternative materials in torque expression would provide further insights into the merits of available wire materials and could guide clinicians in choosing an appropriate material based on the needs of a given patient. Indeed, examining alternative materials to SS may be more clinically relevant given typical archwire sequencing which limits use of SS wires to later stages of care, after much of the pre-adjusted prescription of an appliance has been expressed. Considering known differences in properties between differing wire materials[1,56,57], it is likely that TMA or NiTi wires will generate lower torquing moments than those observed in the present study using SS wires. Indeed this is supported by the previous literature.[22,32,53–55,58] In turn it can be hypothesized NiTi and TMA wires will provide a broader range of twist required for generation of a clinically relevant torquing moment, as has been suggested in the available literature.[22] It can also be speculated ASL clips paired with undersized wires of alternative materials may demonstrate different performance patterns than was observed with undersized SS wires. Similarly, examining materials such as multi-stranded steel would be valuable as some clinicians may select such alternative materials for torque application, *in vivo*.

Finally, repeating multiple torque cycles with one bracket and wire segment will allow assessment of degradation and fatigue over time, which is considerably clinically relevant given the duration typical orthodontic treatment requires to complete. Ideally, conducting such an experiment in conditions simulating the oral environment would be particularly useful.

Chapter 6

6 Conclusions

In vitro examination of torquing moments generated with three ASL, a single PSL, and a single conventionally ligated twin bracket using various SS wire sizes revealed:

1. The PSL and conventionally ligated bracket systems tended to produce significantly greater torquing moments than ASL systems, especially with larger wires and at greater degrees of rotation.
2. Larger wire sizes tended to produce greater torquing moments than smaller wire sizes, regardless of degree of rotation applied and bracket system investigated.
3. Direction of rotation tended to influence torquing moment generated with ASL systems but not the PSL or conventionally ligated systems, with buccal root torque simulations tending to produce significantly greater torquing moments than palatal root torque simulations.
4. The PSL system tended to show smaller engagement angles than ASL or conventionally ligated bracket systems, especially with larger wire sizes.
5. Ligation modality alone does not fully explain the differences in torque expression and engagement angles found between bracket systems, and other aspects of bracket design, such as slot depth and bracket rigidity, likely contribute to these findings.

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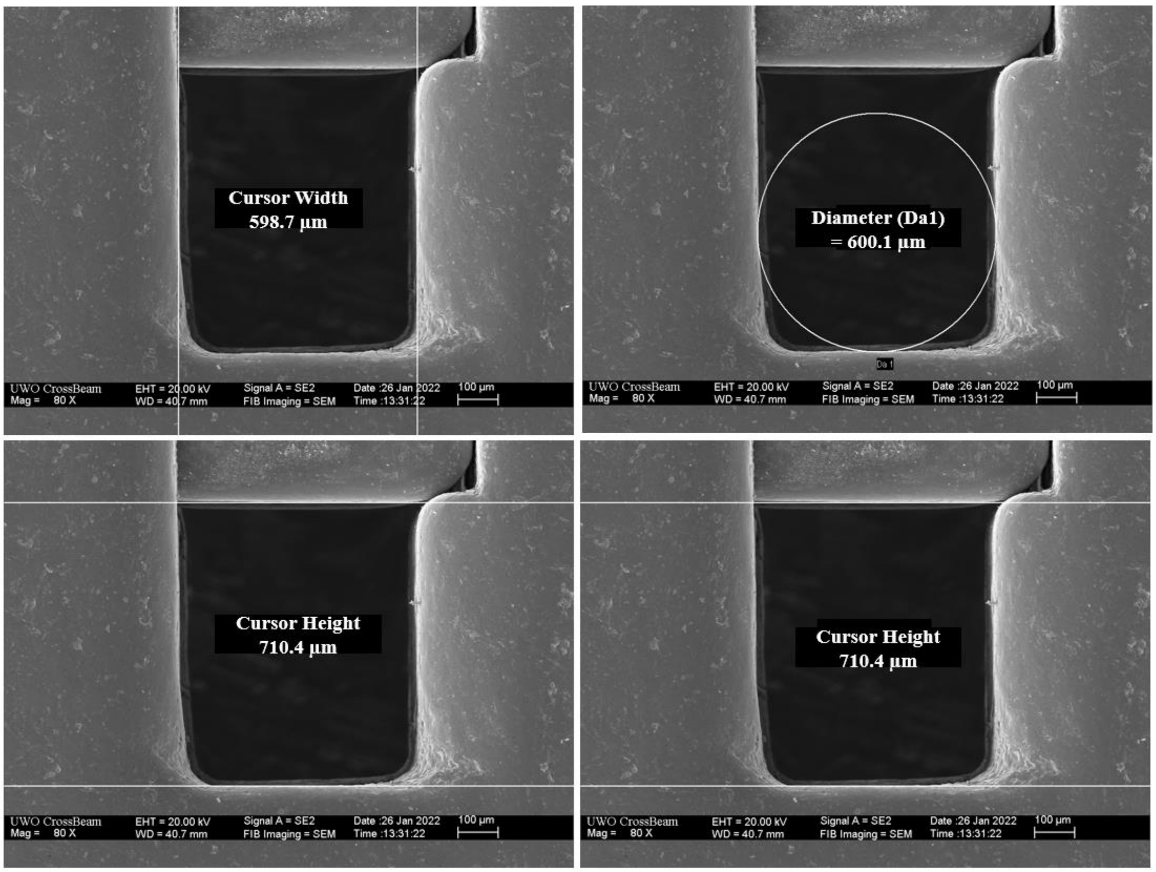
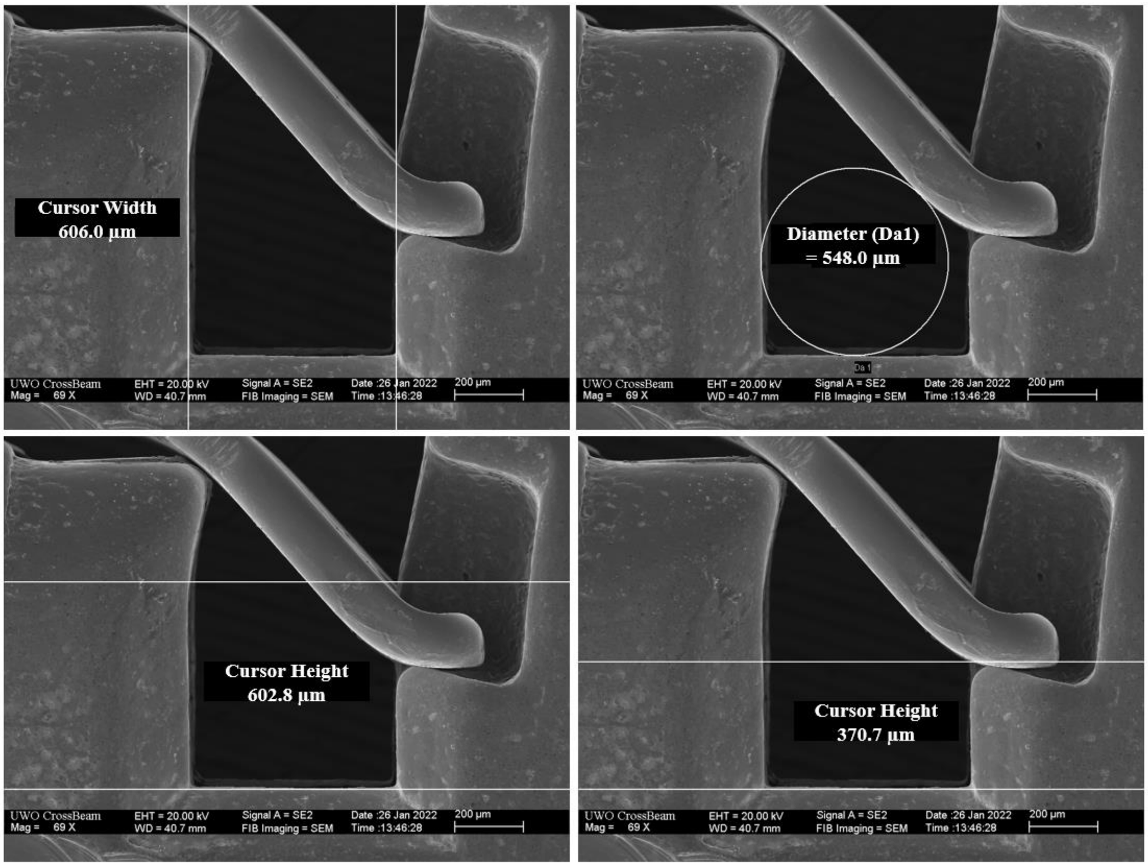
Appendices

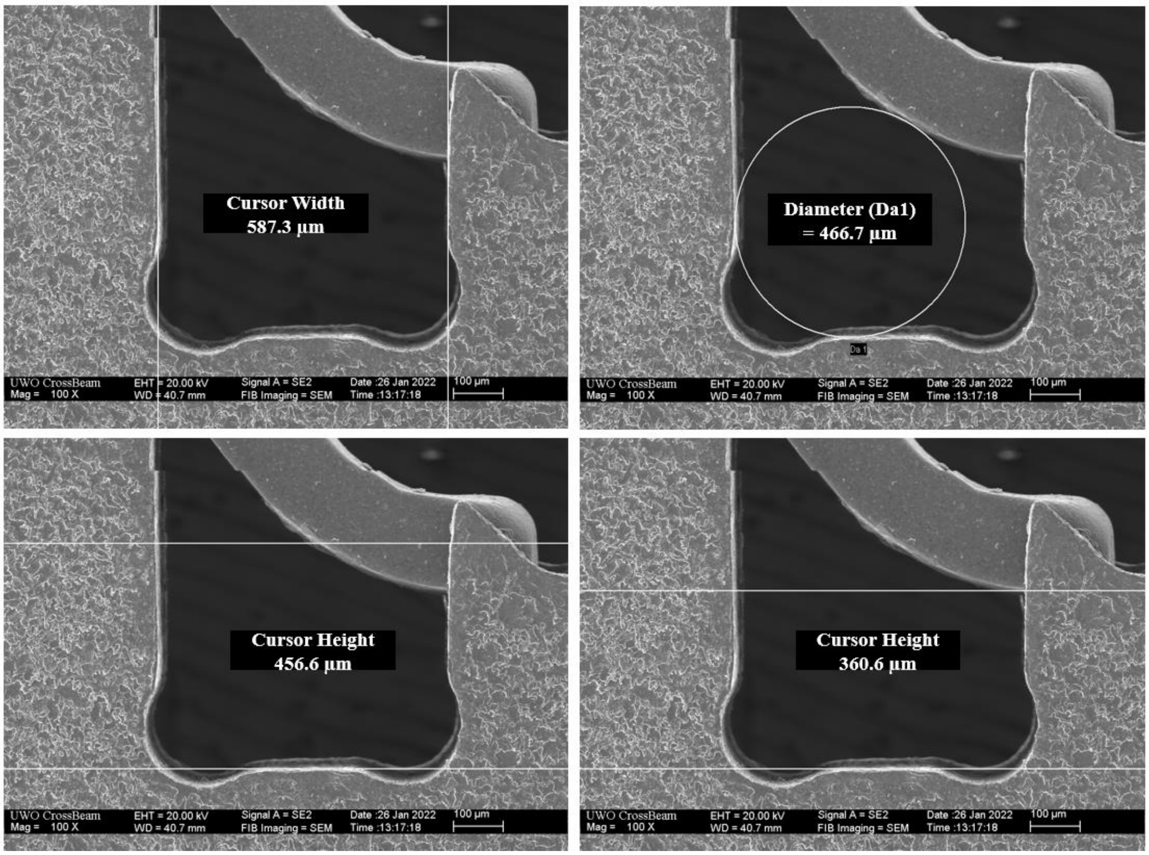
Appendix A: Scanning electron microscopy images of P-Dmn brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 1µm = 0.0397mils



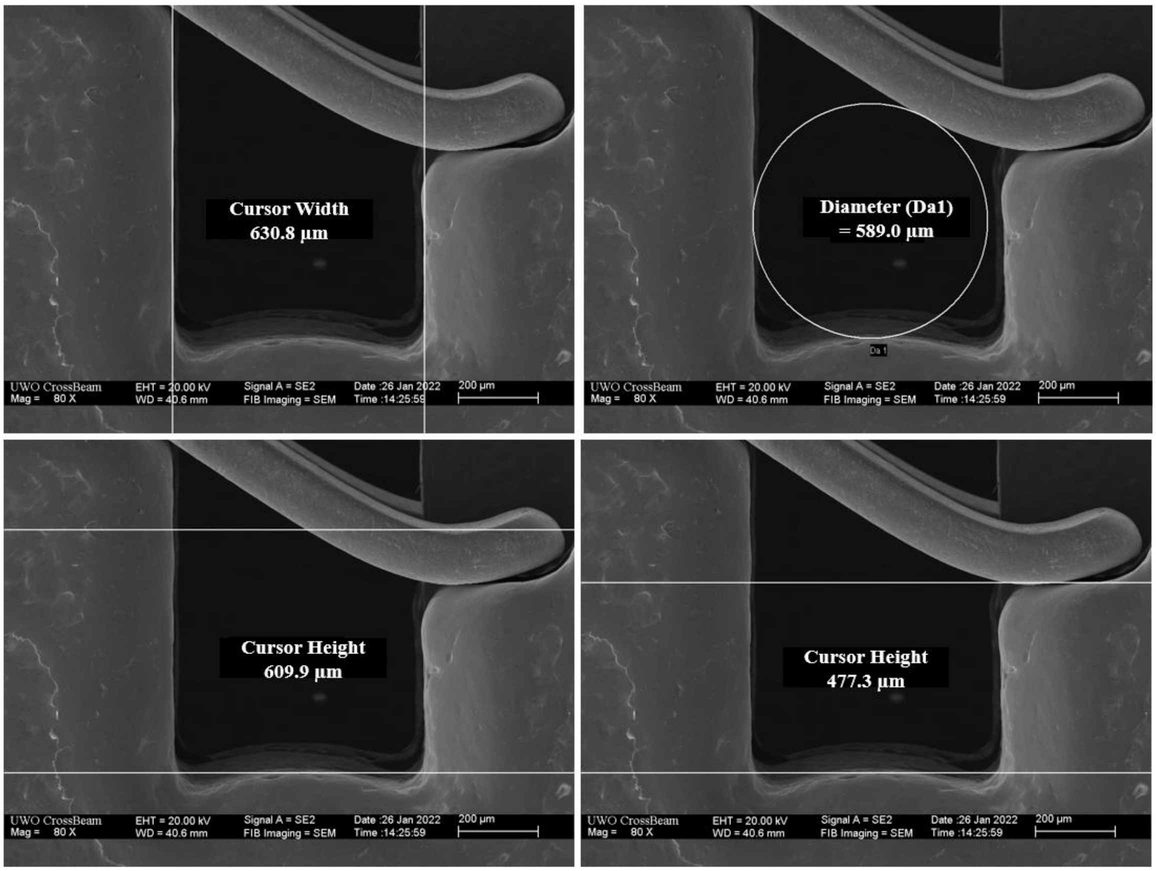
Appendix B: Scanning electron microscopy images of A-Emp brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 1µm = 0.0397mils



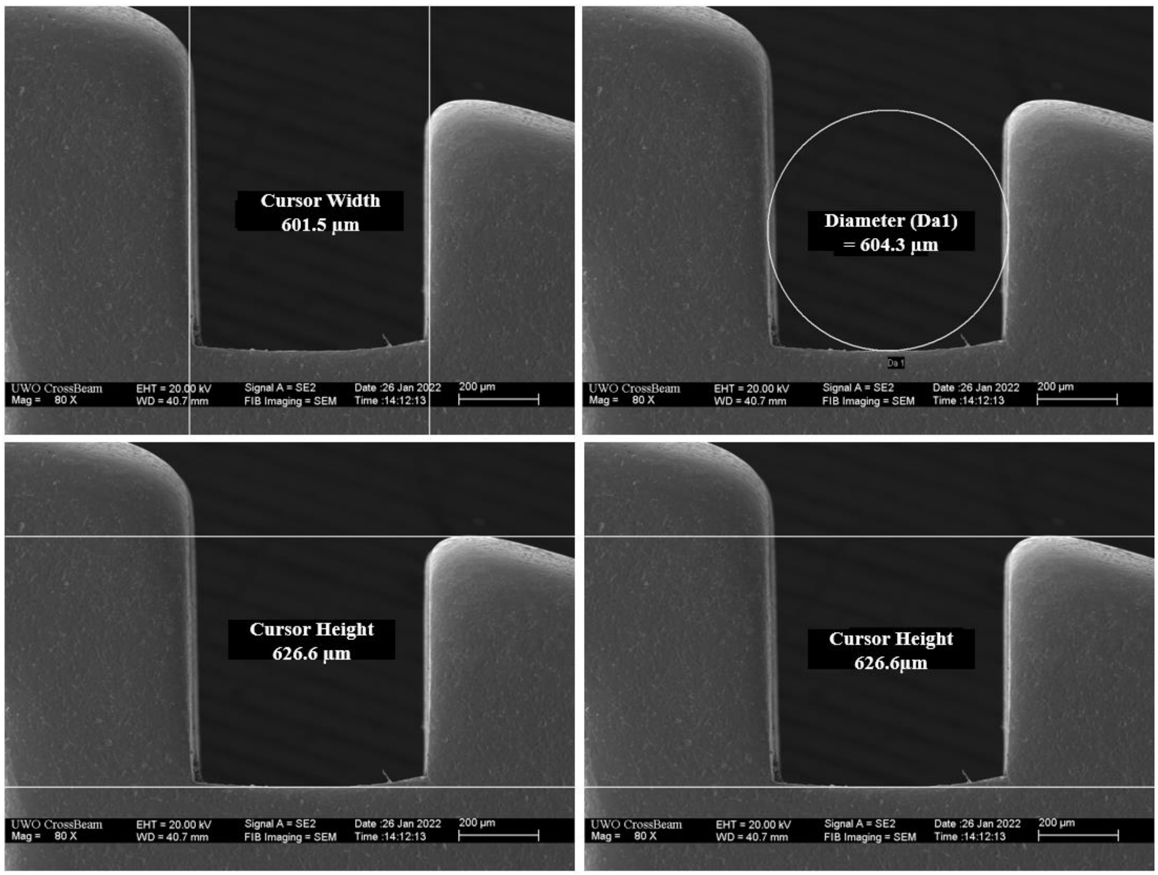
Appendix C: Scanning electron microscopy images of A-Spd brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 1µm = 0.0397mils



Appendix D: Scanning electron microscopy images of A-Vic brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 1µm = 0.0397mils



Appendix E: Scanning electron microscopy images of C-Vic brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 1µm = 0.0397mils



Appendix F: Mean torque values (Nmm) at the nearest approximation to 3-degree intervals of rotation for each tested bracket group with 16 x 22 mil SS wires

	Bracket System									
Angle (º)	P-Dmn(B)	P-Dmn(P)	A-Emp(B)	A-Emp(P)	A-Spd(B)	A-Spd(P)	A-Vic(B)	A-Vic(P)	C-Vic(B)	C-Vic(P)
0	-0.04	-0.04	0.01	-0.06	-0.02	0.04	-0.03	-0.11	-0.02	-0.03
3.04	0.33	0.55	0.98	0.65	0.64	0.27	0.54	0.07	0.66	0.42
5.96	0.45	0.75	1.25	0.90	0.87	0.47	0.76	0.26	0.90	0.63
9	0.56	0.91	1.44	1.05	1.01	0.66	0.87	0.44	1.09	0.83
12.04	0.64	1.09	1.59	1.16	1.16	0.81	0.97	0.61	1.24	1.00
14.96	0.71	1.21	1.70	1.23	1.27	0.97	1.10	0.82	1.39	1.15
18	0.79	1.36	1.76	1.26	1.35	1.10	1.19	0.99	1.52	1.28
21.04	0.91	1.49	1.80	1.23	1.45	1.24	1.29	1.20	1.62	1.38
23.96	1.05	1.58	1.82	1.17	1.52	1.37	1.37	1.37	1.68	1.45
27	1.75	1.70	1.77	1.04	1.57	1.51	1.41	1.46	1.71	1.49
30.04	4.07	3.17	1.74	0.90	2.12	2.07	1.44	1.52	1.72	1.57
32.96	7.00	5.99	1.76	0.91	4.45	3.21	1.48	1.59	1.77	1.76
36	10.45	9.59	2.53	1.20	7.66	5.15	1.56	1.76	2.51	3.13
39.04	14.39	13.75	4.48	2.41	11.19	7.17	1.86	2.05	5.39	5.94
41.96	18.48	17.79	6.88	3.94	14.44	8.86	2.51	2.62	8.60	8.88
45	22.45	21.81	9.17	5.65	17.13	10.07	3.48	3.53	11.67	11.83

Appendix G: Mean torque values (Nmm) at the nearest approximation to 3-degree intervals of rotation for each tested bracket group with 17 x 25 mil SS wires

	Bracket System									
Angle (º)	P-Dmn(B)	P-Dmn(P)	A-Emp(B)	A-Emp(P)	A-Spd(B)	A-Spd(P)	A-Vic(B)	A-Vic(P)	C-Vic(B)	C-Vic(P)
0	0.01	0.02	0.02	0.02	0.00	0.01	0.01	0.01	0.01	0.01
3.04	0.87	0.91	1.84	1.63	1.11	0.41	0.62	0.37	0.44	0.53
5.96	1.24	1.38	2.19	2.13	1.52	0.69	0.87	0.71	0.70	0.78
9	1.74	2.07	2.48	2.18	1.91	0.92	1.14	1.05	0.93	0.98
12.04	2.38	2.78	2.76	2.17	2.20	1.11	1.40	1.31	1.15	1.18
14.96	3.45	3.38	2.93	2.10	2.56	1.19	1.67	1.57	1.39	1.34
18	7.27	6.62	3.23	1.90	4.32	1.55	1.94	1.79	2.78	1.93
21.04	13.39	12.98	6.26	2.08	9.61	3.21	3.58	2.17	8.54	6.87
23.96	19.98	19.50	11.30	5.20	15.80	6.30	7.96	3.27	14.87	13.19
27	26.79	26.21	17.04	8.81	22.29	9.32	13.29	5.67	21.49	20.00
30.04	34.40	33.71	23.57	13.19	29.58	12.35	19.81	9.28	28.93	27.59
32.96	41.19	40.50	29.66	17.26	36.11	15.44	26.16	13.04	35.74	34.60
36	47.83	47.14	35.91	21.37	42.67	18.62	32.79	17.10	42.47	41.73
39.04	54.51	53.76	42.21	25.61	49.14	21.68	39.54	21.68	49.24	48.81
41.96	60.34	59.64	47.93	29.62	54.78	24.13	45.92	26.39	55.23	55.20
45	65.35	64.69	53.07	33.17	59.21	25.71	51.69	30.92	60.54	60.93

Appendix H: Mean torque values (Nmm) at the nearest approximation to 3-degree intervals of rotation for each tested bracket group with 19 x 25 mil SS wires

	Bracket System									
Angle (º)	P-Dmn(B)	P-Dmn(P)	A-Emp(B)	A-Emp(P)	A-Spd(B)	A-Spd(P)	A-Vic(B)	A-Vic(P)	C-Vic(B)	C-Vic(P)
0	0.01	-0.03	-0.04	-0.06	0.02	-0.02	0.02	-0.08	-0.09	-0.02
3.04	0.27	0.52	0.64	0.66	0.71	0.21	0.30	0.28	0.30	0.38
5.96	2.54	2.78	0.86	0.73	1.22	0.57	0.57	0.53	0.57	0.67
9	6.77	6.81	1.75	1.25	2.68	1.92	1.12	0.91	2.03	1.74
12.04	11.41	11.09	5.42	3.79	6.50	5.00	3.69	2.33	6.53	5.78
14.96	16.30	15.25	9.58	7.37	11.22	8.21	7.64	5.13	11.27	10.41
18	21.09	19.93	13.90	11.21	15.96	11.13	11.73	8.57	16.06	15.10
21.04	27.23	27.04	18.75	15.55	21.46	14.12	16.35	12.52	21.53	20.29
23.96	34.28	34.49	24.04	19.63	27.22	16.98	20.93	16.45	27.47	25.94
27	41.39	41.82	30.59	23.87	33.70	19.65	25.99	20.57	34.48	32.46
30.04	49.47	49.83	38.24	29.33	41.17	22.76	32.63	25.54	42.63	40.27
32.96	56.91	56.98	45.23	34.03	47.85	25.91	39.32	30.69	49.77	47.46
36	63.84	63.40	52.28	38.66	54.09	29.12	46.10	36.41	57.04	54.65
39.04	70.49	70.04	59.08	43.31	58.25	31.86	52.99	42.42	64.29	61.62
41.96	76.15	75.31	65.00	47.55	58.00	33.53	59.10	48.35	70.41	67.64
45	80.87	79.83	70.08	51.15	45.38	34.06	64.74	53.80	75.77	72.90

Appendix I: Mean torque values (Nmm) at the nearest approximation to 3-degree intervals of rotation for each tested bracket group with 20 x 20 mil SS wires

	Bracket System									
Angle (º)	P-Dmn(B)	P-Dmn(P)	A-Emp(B)	A-Emp(P)	A-Spd(B)	A-Spd(P)	A-Vic(B)	A-Vic(P)	C-Vic(B)	C-Vic(P)
0	0.00	-0.08	-0.05	-0.09	0.06	-0.02	-0.02	-0.05	-0.04	0.00
3.04	0.24	0.31	0.93	0.37	0.46	0.19	0.53	0.23	0.50	0.35
5.96	1.30	1.12	1.18	0.42	0.57	0.39	0.67	0.45	0.80	0.62
9	5.65	5.94	1.63	1.49	1.34	0.58	0.93	0.69	2.17	1.89
12.04	11.20	11.88	5.53	4.97	5.50	1.28	1.68	1.24	6.85	7.05
14.96	17.40	18.20	10.84	9.02	10.73	2.93	4.22	3.08	12.91	13.07
18	24.12	24.82	16.61	13.30	16.26	5.35	8.21	6.18	19.33	19.63
21.04	31.56	32.13	23.17	18.00	22.71	7.80	13.53	10.42	26.44	26.90
23.96	38.35	38.76	29.39	22.25	28.83	9.86	19.20	14.73	33.05	33.51
27	44.96	45.04	35.45	26.29	34.69	11.71	24.92	18.86	39.54	39.88
30.04	51.73	51.63	41.78	30.39	40.88	13.88	31.25	23.35	46.20	46.49
32.96	57.58	57.13	47.34	34.01	45.65	15.89	37.23	27.55	52.08	52.17
36	63.05	62.41	52.66	37.38	48.28	17.62	43.08	31.66	57.72	57.61
39.04	68.13	67.38	57.80	40.82	43.28	18.29	48.93	36.04	63.16	62.84
41.96	72.46	71.44	62.21	43.88	33.95	17.19	54.30	40.10	67.72	67.30
45	75.95	74.82	65.73	46.45	21.46	16.55	58.89	43.41	71.61	71.01

Curriculum Vitae

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	Dr. Donald S. Moore Award in Occlusal Therapy 2015
	Western Faculty Association Scholarship 2015
	Centennial Orthodontic and Paediatric Dentistry Award 2015
	The Royal College of Dental Surgeons of Ontario Scholarship 2014
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The Royal College of Dental Surgeons of Ontario Scholarship
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