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Torque Expression of Active and Passive Self-Ligating Orthodontic Brackets with Different Archwire Materials

Sidney Mugford, *Western University*

Supervisor: Tassi, Ali, *The University of Western Ontario*

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Abstract

This study aimed to compare torquing moments, engagement angles, and torsional stiffness generated by stainless steel (SS), titanium molybdenum alloy (TMA) and nickel titanium (NiTi) 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. A custom 3D printed testing apparatus was developed to measure torque. In general, the PSL and conventionally ligated systems generated significantly larger torquing moments than ASL systems, especially with stiffer 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 stiffer wires. Torsional stiffness values aligned with the expected modulus of elasticity of the given wire material. In addition to ligation modality, other aspects of bracket design likely contribute to these findings.

Keywords

Orthodontic Brackets, Stainless Steel, Titanium Molybdenum Alloy, Nickel Titanium, Orthodontic Archwires, Torque Expression, Third-Order Tooth Movement, Self-Ligation, Engagement Angle, Torsional Stiffness

Summary for Lay Audience

Patients often seek orthodontic care to achieve a straighter, more esthetic smile, through which the orthodontist must properly position the teeth in the mouth in all three planes of space. Braces, or orthodontic brackets, are commonly used in conjunction with wires to move teeth into their ideal orientation. A common tooth movement achieved with braces, called "torque", involves changing the angulation of teeth by moving the tooth roots toward or away from the lips or cheeks.

Three basic types of orthodontic brackets are available today, which differ in how the wire is secured to the bracket, or the "ligation method". These different types of ligation methods are each suggested to have different advantages, with one type (active self-ligation, or ASL) suggested to add torque to teeth more efficiently than other methods. There are also different wire materials which are commonly used throughout orthodontic treatment, which vary in terms of their properties such as flexibility and stiffness. To test which bracket-wire combination produces torque most effectively, five different bracket systems representing the three ligation methods were tested by twisting brackets from -15 to 45 degrees, clockwise and counterclockwise, around a section of orthodontic wire and measuring resulting torquing moments. This was repeated in two directions (root towards the lips, and root away from the lips). Resulting torque moments were compared between different bracket types, wire materials, directions of twist, and system stiffness, to existing literature to determine if one ligation method was superior in producing torque.

In general, with higher degrees of twist, torquing moments increased for all bracket systems and wires tested. For a given degree of twist, torquing moments tended to be higher for stiffer wires than more flexible wires. In 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. System stiffness values were lower in the ASL groups as well. These findings are likely the result of not only ligation modality, but other aspects of bracket design as well.

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List of Abbreviations

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
A-NiTi	Austenitic nickel titanium
ANOVA	Analysis of variance
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
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
GPa	Gigapascal
in	Inch
MBT	McLaughlin, Bennett, Trevisi prescription
M/F	Moment-to-force ratio
MIM	Metal injection molding
mm	Millimeter

M-NiTi	Martensitic nickel titanium
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
PDL	Periodontal ligament
PSL	Passive self-ligation
SD	Standard deviation
SEM	Scanning electron microscopy
SL	Self-ligation
SS	Stainless Steel
SWA	Straight Wire Appliance
TMA	Titanium molybdenum alloy

Chapter 1

1 Literature Review

1.1 Introduction

In today's society, orthodontic treatment is commonly sought for patients of all ages. Irregular, misaligned, or protruding teeth can cause many issues for patients, including psychosocial problems, a greater susceptibly to trauma, tooth decay or periodontal disease, and problems with oral function.[1] In earlier years orthodontists viewed their goals as straightening teeth and correcting malocclusions, but today's focus is a broader mission to improve both facial and dental appearances while maintaining proper relationships of teeth to one another.[2] To achieve both these esthetic and functional goals, the American Board of Orthodontics[3] has set out eight criteria which should be achieved with orthodontic treatment: ideal tooth alignment, ideal marginal ridge relationship, proper buccolingual inclination, proper occlusal relationship, adequate occlusal contacts, proper overjet, ideal interproximal contacts, and proper root angulation. Proper buccolingual inclination, also known as crown inclination, allows for proper overbite and posterior occlusion.[4] Proper buccolingual inclination of the anterior teeth in particular establishes an esthetic smile line, proper anterior guidance, and a Class I molar and canine relationship, while improperly inclined teeth can affect arch length and space requirements.[5] Achieving proper buccolingual inclination, or more simply torque, of the teeth is clearly important to accomplishing the goals of orthodontic treatment. It is essential that the orthodontist has a complete understanding of the biomechanical principles of tooth movement, as well as the factors which influence torque expression in contemporary orthodontic appliances.

1.2 Principles of Tooth Movement

Forces applied to teeth, bones, and the periodontium by the orthodontist are primarily how dentofacial changes are achieved. Hence, Newtonian mechanics applied to a biological system and physics make up the scientific basis of orthodontics.[6] As the teeth and their support structures respond to applied forces, a biological cascade of events occurs which ultimately results in the tooth moving through its surrounding bone.[7] The periodontal ligament (PDL) mediates the bony response to forces, therefore tooth movement is primarily a phenomenon of the PDL.[8] Biomechanics is the science of mechanics in relation to biological systems, and an understanding of the application of biomechanical concepts forms the basis of orthodontic treatment.[7]

A force is defined as a vector with both a magnitude and direction. Commonly forces are expressed in units of Newtons (N), however in orthodontics forces are usually expressed in grams (g).[9] The magnitude of a vector represents its size, while the direction is described by its point of origin and line of action, and these are represented by arrows.[7] The length of the arrow is proportional to the magnitude of the force, the origin of the arrow represents the point of application of the force, and the arrowhead indicates the direction of force.[10] When a force is applied to a free body, an object in which all its mass is centered, and said force is applied through the body's center, the body will move in the same direction as the line of force. This is known as translational movement.[9]

A tooth is not a free body, as it is restrained by its supporting periodontal structures. The center of resistance of a tooth is analogous to the center of mass of a free body. The location of the center of resistance is dependent upon the tooth's root morphology, root length, number of roots, as well as the height of alveolar bone surrounding the tooth. If a force acts at the center of resistance, the tooth will translate or move in the same direction as that force.[9] The center of resistance for a single-rooted tooth is usually located one-half to one-third of the root length apical to the alveolar crest. For a multi-rooted tooth, the center of resistance is usually found between the roots, 1 to 2 millimeters (mm) apical to the furcation.[10] Forces most commonly act on teeth through brackets, which are bonded to the crown, such that it limits the opportunities for the force to pass through the center of resistance. When a force acts away from the center of resistance, the resulting movement is rotation.[9]

The potential for rotation is measured as a moment, of which the magnitude is equal to the force multiplied by the perpendicular distance of the line of action of the force to the center of resistance. The moment of the force is reported in units of gram-millimeters (g-mm), assuming the force is measured in grams and the distance in millimeters. The direction of the moment can be determined by continuing the line of action of the force around the center of resistance.[10] Total tooth movement that occurs from a force acting away from the center of resistance is a combination of rotation and translation. The tooth will be displaced, or translate, in the direction of the force, while also rotating around its center of resistance. This results in tipping of the tooth as it moves.

Another force system which can be applied to a tooth is a couple. A couple is defined as two forces equal in magnitude and opposite in direction, and separated by a distance (having different lines of action).[1,9] By applying two forces in this way, the translatory effect is cancelled out because they are equal and opposite, resulting in a pure moment and pure rotation of the tooth.

1.3 Classifications of Tooth Movement

There are essentially four categories of tooth movement: tipping, translation, root movement, and rotation. These movements can occur in all three planes of space, but for this investigation the focus will be on tooth movement in the sagittal plane. Each type of movement is the result of a different force and moment applied to the tooth, which is called the moment-to-force (M/F) ratio[7]. Tipping is the result of greater movement of the crown than of the root, with the center of rotation being located apical to the center of resistance of the tooth. If the center of rotation is at the apex of the root, controlled tipping will occur. However, if the center of rotation is located between the center of resistance and the apex the tooth will experience uncontrolled tipping. Uncontrolled tipping is the simplest type of tooth movement to produce, and causes the crown and the root to move in opposite directions, which is often undesirable (Figure 1A).[10] The M/F ratio is typically between 0:1 and 5:1. When the M/F ratio is increased to 7:1 controlled tipping will occur, in which the crown of the tooth will move in the direction of the force while the root apex maintains its position (Figure 1B).[11]

When the M/F ratio is increased to 10:1 translation will occur, in which the crown and the root apex will both move the same distance and in the same direction. This is also known as bodily movement (Figure 1C). The center of rotation is infinitely far away from the center of resistance. Changing the tooth's axial inclination while maintaining the crown position is known as root movement, and here the center of rotation is located at the crown of the tooth. The M/F ratio required is 12:1 or greater.[11] Root movement in orthodontics is often commonly called torque (Figure 1D).[7] As discussed above, rotation occurs when only a couple is applied to a tooth.[12]

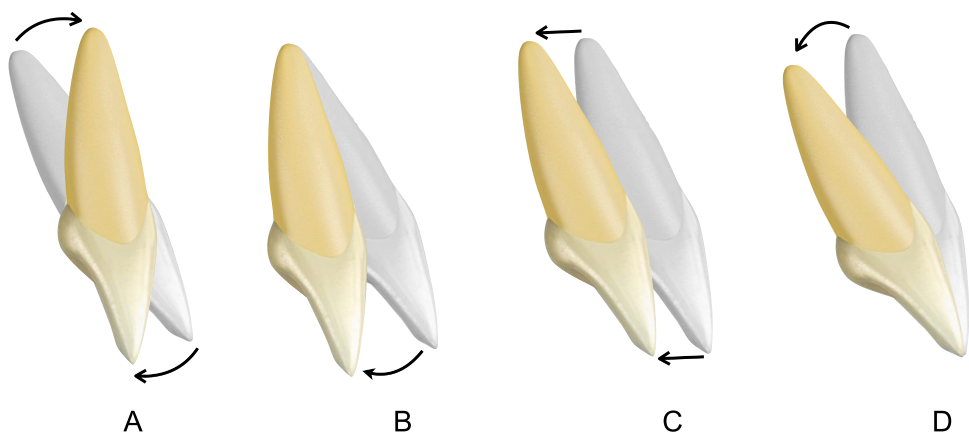


Figure 1: Categories of orthodontic tooth movement, demonstrating uncontrolled tipping (A), controlled tipping (B), translation (C), and root movement (D)

Tooth movement can also be described in orthodontics by first-, second- and third-order movements, which were coined by Edward Angle when he designed his appliance. First-order movement is in the faciolingual dimension and is also referred to as in-out movement. Second-order movement, also known as tip, refers to mesiodistal movement of the crown or root. And lastly, third-order movement, or torque, refers to root movement.[1]

1.4 Contemporary Orthodontic Appliances

There are several different appliance options available to provide the optimal force system needed to move teeth, but the classic approach is fixed appliances, which use brackets bonded to teeth in combination with archwires. The fixed appliances used today are based on Edward Angle's appliance designs from the early 20th century. Angle introduced the Edgewise appliance in 1928, which consisted of a horizontal rectangular slot, 0.022 x 0.028-in in dimension, and utilized archwires fabricated from precious metals, such as gold alloys, of the same dimensions. This provided good control of both the crown and root in all three planes of space but lacked the ability to carry out any sliding mechanics such as those used to close extraction spaces.[2]

As gold alloy archwires were replaced with steel, the edgewise appliance was redesigned to optimize the bracket slot dimensions for use with steel. Due to the fact that steel is much stiffer than gold of the same dimension, it was advocated a that reduction in slot size from 0.022-in to 0.018-in should be made. Despite the smaller dimensions, slightly greater forces were able to be produced with full dimension steel archwires when compared with the original edgewise appliance. At this time, it was also noted that there could be advantages in using undersized archwires in the original edgewise brackets. The amount of friction as teeth slide along the archwire when spaces were being closed was reduced, but this then presented the disadvantage of later being unable to produce sufficient torque.[1]

In an attempt to use steel archwires in the 0.022-in bracket slot to produce torque, 0.021-in archwires were proposed, but springiness and range in torsion were limited such that effective torque with the archwire was essentially impossible. Smaller rectangular archwires, such as 0.019 x 0.025-in, could be used with exaggerated inclinations, or third-order bends, but it was found that torquing auxiliaries were still required. This opened the door for archwires of newer materials to help overcome the major problems associated with the original edgewise slot size, which will be discussed further below.[1]

Prior to the 1970s the same bracket was used on each tooth, and first-, second-, and third-order bends were necessary to compensate for the differences between tooth anatomy. At this time Lawrence Andrews developed bracket modifications that created different brackets for each tooth, eliminating the repetitive bends needed in archwires. This was termed the Straight Wire Appliance (SWA), and was key to improving the efficiency of the original edgewise appliance.[13] Using the 120 non-orthodontic models Andrews used to develop "The six keys to normal occlusion"[4] he converted this knowledge into a "preadjusted" appliance, with first-, second-, and third-order values built into each bracket.[14] First-order, or in-out bends, were compensated by varying the thickness of the base of the bracket depending on which tooth the bracket would be bonded to, to make up for the variations in the contour of the buccal surfaces of individual teeth. Second-order, or tip bends, were addressed by angling the bracket slot relative to the long axis of the tooth.[1] And finally, to reduce the need for third-order, or torque bends, the bracket base angle was modified from 90 degrees to different levels of acute or obtuse angles, referred to as torque in base.[14] These built in angulation and torque values are what are now known as the "appliance prescription" of modern orthodontic brackets.[1]

Figure 2: A selection of contemporary orthodontic brackets, including metal and ceramic, as well as conventionally ligated and self-ligating varieties

In today's market there exists many different variations of the SWA. Brackets can be fabricated using different materials, most commonly metal brackets are composed of stainless steel, but there also exists brackets made of titanium as an alternative to steel, to be used in patients who present with a nickel allergy. Non-metallic materials, such as plastics, ceramics, and plastic composites are also used to fabricate brackets for a more esthetic appearance. In addition to the different materials available to fabricate brackets, they may also differ in the method in which the archwire is held in the bracket slot.

Traditional twin brackets used stainless steel ligature wires tied around the tie wings of the bracket to hold the archwire in the slot. This was a time-consuming procedure, and has largely been replaced by the introduction of the elastomeric modules in the 1970s. Brackets with built-in ligation mechanisms, known as self-ligating (SL) brackets are also available.[14] The first SL brackets were introduced in the 1930s but they did not begin to gain popularity until the 1990s. Self-ligating brackets can be further divided into two categories: active SL brackets, composed of a sliding spring clip which potentially places an active force on the archwire by encroaching on the bracket slot from the labial aspect; and passive SL brackets, which have a slide door that is passive and has no ability to invade the bracket slot.[15]

1.5 Archwire Materials

Currently there exists many different orthodontic archwire alloys, each that exhibit a wide variety of properties, available for the orthodontist to select to best meet the demands of the clinical situation. It is important that the orthodontist has a thorough understanding of the mechanical properties of the wire to be able to provide ideal and predictable treatment results.[16] The most commonly used alloys include stainless steel (SS), nickel-titanium (NiTi), and titanium-molybdenum alloy (TMA or beta-titanium), and will be the focus of further discussion.

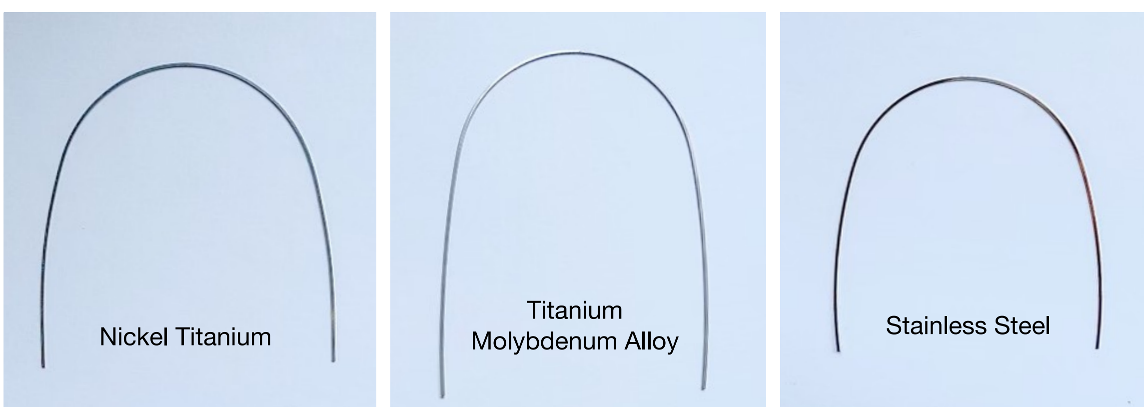


Figure 3: Examples of orthodontic archwires in varying arch forms

Stainless steel archwires are typically made of alloys which contain 17-25% chromium, 8-25% nickel, with the remaining balance being iron.[17] Typically, the formulation is referred to as 18-8 SS as it contains 18% chromium and 8% nickel, and this formulation provides corrosion resistance. The mechanical properties of SS wires can be controlled through the amount of annealing and cold working during the manufacturing process. Wires which are fully annealed will be dead soft and highly formable, or the wires can be hardened by cold working.[1] Stainless steel has a high modulus of elasticity and associated high stiffness, which can be seen as either an advantage or disadvantage, depending upon the stage of treatment and necessary tooth movements. High stiffness requires smaller diameter wire sizes for teeth which are displaced, which can lead to a loss of control during tooth movement. Yet the high stiffness is desirable in the resistance of deformation caused by extra-oral and intra-oral tractional forces. Stainless steel also has low springback and stored energy, as indicated by its yield strength to elastic modulus ratio. This implies that SS wires require frequent activation and archwire changes, as the wires produce high forces which dissipate over short periods of time.[16]

Nickel-titanium is composed of a 50:50 nickel-titanium composition, and the material was first introduced to orthodontics in the 1970s. It can exist in two different crystal structures at intraoral temperatures, and the transition between the two structures is fully reversible. This gives NiTi properties not seen in any other archwire, including superelasticity.[1] When the alloy exists in the martensitic phase it exhibits a low stiffness (elastic modulus of 31-35 GPa) but higher strength (1.4-1.7 GPa) when compared to the austenitic phase, which has an elastic modulus of 84-98 GPa, giving it a higher stiffness, and a lower strength of approximately 0.84 GPa. For perspective, it should be noted that SS has a modulus of elasticity of 200 GPa and ultimate strength of 2.1 GPa. In the austenitic alloy (A-NiTi), both the A-NiTi and martensitic (M-NiTi) phases play a role during mechanical deformation. The alloy begins in the A-NiTi phase and then will undergo a stress-induced phase transformation to the M-NiTi, which creates a plateau effect when observing a force-displacement plot of the alloy, often referred to as pseudoelasticity.[17] These unique mechanical properties give NiTi good springback and flexibility, which increases clinical efficiencies since fewer archwire changes or activations are required, thereby reducing clinical chair time requirements. Also, when a rectangular NiTi wire is inserted early in treatment it can accomplish leveling, correction of rotations, and torquing simultaneously. Nickel titanium does exhibit disadvantages as well, such as its low stiffness which does not offer adequate stability required later in treatment, and its poor formability which makes bending loops or stops very difficult.[16]

Titanium-molybdenum alloy, originally known as beta-titanium, was introduced to the orthodontic market in the 1980s by Charles Burstone. His objectives were to introduce a new alloy which would have large springback, would be highly formable, and have a lower stiffness than SS.[18] The alloy is composed of 80% titanium, 11.5% molybdenum, 6% zirconium, and 4.5% tin.[17] It represents a mid-range of stiffness in comparison to SS and NiTi, with a modulus of elasticity that ranges from 99-122 GPa. Advantages of TMA in comparison to NiTi is that it has good formability and weldability. In comparison to SS, TMA offers gentler forces per unit of deflection, has a higher springback (twice that of SS), and more range. The drawback of TMA is its high coefficient of friction, which limits its ability to slide teeth for closing spaces or retraction. Stainless steel remains the wire of choice for sliding mechanics.[16,17]

In orthodontics, the ideal archwire material would have to possess many characteristics, such as good formability, weldability, resilience, and springback. Ideally, it would also have a low coefficient of friction, poor biohostability, and be esthetic. From the above discussion, it is clear that there is not one single alloy that will offer all the desirable properties. The orthodontist must evaluate the needs throughout the course of treatment to select the most appropriate archwire alloy, shape, and size. As different stages of treatment have differing demands, it is important that the orthodontist has an understanding of the mechanical properties of archwires and their clinical applications.[16,17]

Table 1: Comparison of clinical characteristics of orthodontic archwires

Wire	Springback	Stiffness	Formability	Stored Energy	Joinability	Friction
NiTi	High	Low	Poor	High	Not joinable	Low-moderate
TMA	Moderate	Moderate	Moderate	Moderate	Welded	High
SS	Low	High	Good	Low	Soldered Welded	Low

1.6 Stages of Comprehensive Orthodontic Treatment

Typically, orthodontic treatment can be divided into three phases, each with differing techniques and wire that may be utilized. According to Proffit[1] and McLaughlin's[19] classic orthodontic texts the major stages of orthodontic treatment are broken down into three categories: (1) levelling and aligning, (2), space closure and correction of molar relationships, and (3) finishing.

The goal of levelling and aligning is to complete the tooth movements needed such that a large rectangular archwire can be passively placed into the bracket slots. In working with a 0.022-in bracket slot, this wire would be a 0.019 x 0.025-in SS wire. The ideal archwires for this stage of treatment would have good strength, excellent springiness, and a long range of action. The wires should also deliver light, continuous forces of approximately 50 g, as this will produce efficient tipping of the teeth. Root movement is to be avoid during this stage, as it slows down the alignment and increases the possibility of root resorption. For the reasons listed above, round NiTi archwires are the wires of choice for initial alignment. Leveling of the arch should also be completed with round archwires, to avoid unnecessary torque, particularly on the on mandibular anterior teeth, that would occur with the engagement of a rectangular archwire. Either preformed NiTi wires with bite-opening curves, or round SS wires with curves and/or step bends, can be used to correct the vertical discrepancies and produce a level dental arch.

Once leveling and aligning has been completed and a large rectangular archwire can be passively placed, the second stage of space closure and molar relationship correction can begin. The two major ways to accomplish space closure are sliding and closing loop mechanics. Sliding mechanics involves sliding teeth along the archwire, in which there can be significant resistance to sliding in the system depending on the combination of bracket system and archwire being used. The advantage with pre-adjusted appliances is that the bracket prescription is automatically expressed such that root paralleling movements will occur. Meling et al[20] noted that one can expect 50 g of frictional force to be generated by each elastomeric module used to ligate the archwire. It has been shown that SL bracket systems produce less frictional force, and therefore could be more advantageous in the use of sliding mechanics.[21] Closing loop mechanics involve tying teeth tightly to archwire segments and using a spring to move the segments together. This does not involve any frictional resistance but does require more work and complexity, as it is necessary to design and adjust the spring, or closing loop, to generate root paralleling moments which must also be proportional to the force used to close the space. Both methods utilize large rectangular archwires, usually of TMA or SS for their formability and stiffness.

Correction of molar relationships can occur with the use of many different auxiliaries such as extra-oral traction, functional appliances, inter-arch elastics, or through differential anteroposterior movement of the upper and lower teeth. The choice of which method to use will be determined by the present malocclusion, the growth potential, as well as the orthodontist preference. The use of full sized rectangular archwires during this stage will allow the bracket prescription to express, as the wire engages the bracket slots. At the completion of this second stage of treatment, the teeth will be well aligned with their roots reasonably parallel, extraction spaces will be closed, a Class I buccal segment will be achieved, and vertical discrepancies of deep bite or open bite will be resolved.[19]

Finishing, the final stage of treatment, prior to the introduction of the SWA was much more involved due to multiple bends being necessary to achieve the best possible position of teeth and esthetics. The built-in features of preadjusted system have simplified this stage. There are, however, still adjustments for individual teeth that must be made, such as those required to level marginal ridges, obtain appropriate in-out relationships, and to overcome any discrepancies that have been introduced due to improper bracket placement. The addition of third-order bends to correct torque are also often needed, although this will depend on the bracket system being utilized. In an 0.018-in bracket slot with a full sized 0.017 x 0.025-in SS archwire, full expression of the bracket prescription torque can be expected. However, in a 0.022-in bracket slot with a full-sized SS archwire (0.021 x 0.025-in), the wire is too stiff such that effective torque cannot occur. Full dimension NiTi, TMA or braided SS wires could be used to express the built-in torque, or auxiliaries with smaller dimension archwires may be necessary to achieve proper torque.[1,19]

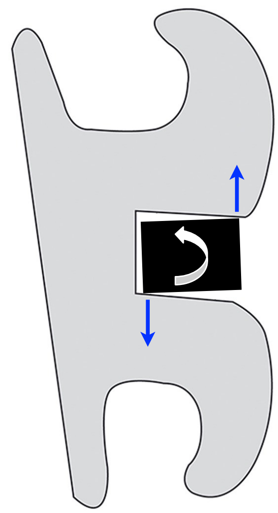


Figure 4: Cross section of a wire twisting within a bracket slot to produce torque, as viewed in the sagittal plane. The black rectangle represents a wire engaged in the slot, the white arrow represents the direction of torsion of the wire, and the blue arrows represent contact points of the wire against the slot walls producing a couple that would generate torque

1.7 Torque Values in Orthodontics

All tooth movements are based on the observation that when prolonged light forces are applied to a tooth, it will move as the bone around the tooth remodels. The response of the bone is mediated by the PDL in such a manner that tooth movement is ultimately a PDL phenomenon. The force distribution within the PDL, and therefore the pressure, will differ depending on the tooth movement that is being elicited. It is necessary to understand the optimal force level for a given tooth movement to occur in a biologically sound manner, to produce the movement in such a way that frontal resorption occurs rather than undermining resorption.[1] In the literature, a clinically effective range of torque can be considered between 5 and 20 Nmm.[22–24]

When a torquing movement is applied to a tooth the centre of rotation is located at the crown, as noted earlier, and the stress levels within the periodontal ligament are greatest at the apex of the root.[7] This concentration of stresses can be up to four-times that compared with translatory tooth movements and may lead to undermining resorption. If the forces are uncontrolled or too high, the risk of apical root resorption increases. Other factors that may influence root resorption and torque aside from force magnitude are the duration of force application, force direction, treatment mechanics and the duration of treatment.[25] The root apex of a tooth is also less mineralized and softer, and contains less Sharpey's fibres which act to buffer stress, making this area more susceptible to root resorption.[26]

While there also exists a genetic component of susceptibility of root resorption, treatment factors are of interest to clinicians as they can be modified to limit the stress, and as such, forces and root resorption has been widely studied. Casa et al [27] investigated the effects of lingual root torque on premolars using scanning electron microscopy and found the severity of root resorption, in both width and depth of lacunae, increased with the magnitude as well as the duration of forces. They also found that areas of severe resorptions were found at the apex, in addition to the bucco-cervical regions which were more discrete. Bartley at al[26] were also in agreement, and in their study on buccal root torque of maxillary first premolars found that higher magnitudes of force caused more root resorption, particularly in the apical region. They applied conservative levels of torque, at 2.5 degrees and 15 degrees, which amounted to force levels of 0.48 and 2.85 Nmm respectively. The judicious clinician must have a thorough understanding of the factors which influence torque expression, so that treatment can be delivered in such a way that minimizes unwanted side effects while maximizing efficiency.

1.8 Engagement Angle

When a square or rectangular archwire is inserted into a bracket slot, if the archwire dimensions are smaller than that of the bracket slot, the wire will have some freedom to rotate within the slot before contacting the walls. Once the archwire has engaged the walls of the bracket slot, it will undergo torsion and be capable of producing a couple to produce torque. This angle at which the clearance between the archwire and bracket first disappears is known as the contact angle, engagement angle, or slop (Figure 5). Engagement angle is influenced by both the bracket slot and archwire dimensions.[28] As the archwire dimensions increase or the bracket slot dimensions decrease, the engagement angle diminishes which will result in increased torque expression.[29]

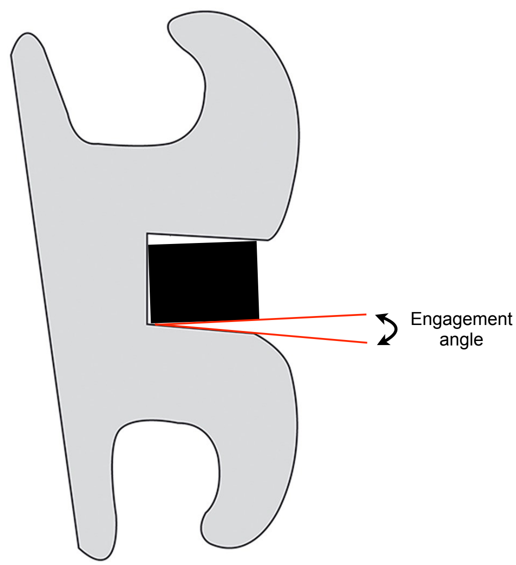


Figure 5: Visualization of the engagement angle between an orthodontic bracket and archwire

To determine the amount of slop, or play, in a given bracket-wire combination, theoretical calculations have been completed. The engagement angle of a full sized 0.019 x 0.025-in archwire in a 0.022-in bracket slot is estimated to be approximately 8 degrees, while the engagement angle of a 0.018-in bracket slot with a full sized archwire of 0.017 x 0.025-in is approximately 4 degrees.[30] Clinical studies have found engagement angles tend to be larger than the theoretical calculations.[31–35] For example, Meling at al[32] demonstrated that one can expect the amount of play in a system of a 0.018-in bracket slot and a 0.017 x 0.025-in archwire to range from 8 to 11 degrees. Joch et al[35] found nearly 12 degrees of play in their study on the engagement angle of 0.019 x 0.025-in archwires in 0.022-in bracket slots. The reasoning for these differences in theoretical and studied engagement angles will be discussed further below.

1.9 Factors Influencing Torque Expression

According to Proffit[1], there are three factors which determine the amount of torque one can expect to be expressed when a square or rectangular archwire is engaged in a rectangular bracket slot. The first is the properties of the archwire itself, such as the size, shape, and material. The second is the inclination of the bracket slot relative to the archwire, which can be influenced by the bracket prescription as well as the anatomy of the tooth. The last is the tightness of the fit between the archwire and the bracket, which can be influenced by numerous factors, such as the design of the bracket, including the slot dimensions and bracket material, as well as the ligation method. Each of these factors can in turn be influenced by numerous other factors, which will be discussed in detail below.

1.9.1 Archwire Shape and Dimension

For a given bracket to express torque, the archwire must engage the bracket slot at two points on opposite edges to create the moment of a couple.[1] The fit between the archwire and the bracket slot is known as play, slop, or the engagement angle, which represents the amount of rotation in degrees that a rectangular or square wire must be twisted to engage the slot and generate torque.[36] Generally, increasing the cross-sectional dimensions of a wire demonstrates a reduction in the amount of play[33], resulting in higher torque magnitudes. As well, square archwires generate smaller torque magnitudes when they are compared to rectangular archwires.[37]

Determining the amount of torsional play that exists in different archwire-bracket slot dimension combinations are possible in theory, however studies have shown that the theoretical calculations often to not match clinical situations, due to differences in reported dimensions of both archwires and bracket slots. Archwires dimensions are more commonly undersized when compared with their reported dimensions, however oversized wires have also been observed. Variations in archwires also exist in the edge bevel, or rounding of the archwire corners that occurs during manufacturing.[38] A wire with a high degree of bevel will have more play than expected, resulting in a poorer fit in the bracket slot which may result in less control during tooth movement.[32]

1.9.2 Archwire Material

An overview of the properties of contemporary archwire materials was discussed above. That which relates most to torque expression is the torsional stiffness of the material. Torsional stiffness is defined as the measure of a rectangular profile's resistance to torsion and is dependent upon the mechanical characteristics and geometrical dimensions of the twisted profile.[39] Low stiffness provides the ability to apply lower forces, which are more constant over time as deactivation occurs, in addition to greater ease and accuracy in applying a given force. High stiffness provides the advantage of resisting deformation.[16] Of the commonly used archwire materials in orthodontics, SS has the highest stiffness, followed by TMA, with NiTi having the lowest stiffness.[40]

It has been stated that materials of low stiffness, such as NiTi and TMA, are ineffective at generating torquing moments in bracket slots.[41] Generally, at low degrees of torque when the wire is not engaged in the bracket slot, torque expression between archwire materials shows little to no significant differences.[42–44] As the wire engages the slot, SS is shown to produce the highest torquing moments when compared to other archwire materials, and NiTi shows the lowest.[42,44] In other words, SS wires are capable of producing torque at smaller angles than NiTi wires of the same dimension.[45] Quantitatively, Archambault et al[42] found that for angles of twist greater than 24 degrees SS produces 2.5 to 3 times greater torque moments compared to NiTi, and 1.5 to 2 times that of TMA. Similarly, Huang et al[43] found that theoretically changing the wire material from NiTi to TMA increases torquing moments by 34%, while changing to SS shows a 220% increase.

As previously noted, clinically effective torque values have been found to be between 5 and 20 Nmm. Due to the high stiffness of SS, there is only a small range of wire twist that will provide torquing moments in this physiologic range, especially as wire size increases. Clinically this can make controlling torquing forces difficult and may place unnecessarily high forces on the dentition if the clinician is not precise. The tendency for SS to produce the greatest torquing moments is not always considered an advantage, especially in clinical scenarios.[32] In contrast, the trend for NiTi to generate the lowest torquing moments is also not necessarily an advantage to clinical practices. A significant amount of twist may be needed to produce torquing moments in the physiologic range, which can prove to be difficult in a wire which does not have good formability.[33] However, manufacturers have recognized this downfall and have produced NiTi wires with preadjusted torque to aid the clinician with torque expression.[42]

1.9.3 Bracket Prescription

As described above, the bracket prescription is a set of modifications to the bracket slot and base design that allows for predetermined movements to be programmed into the bracket, eliminating the need for many repetitive wire bends. In terms of torque, it is achieved by either cutting the bracket slot at an angle, or forming the bracket base such that it is at an angle, known as torque-in-face or torque-in-base, respectively. Commonly used bracket prescriptions today include Roth, and McLaughlin, Bennett, Trevisi (MBT), but a large range of options exist. When considering the maxillary central incisor, the Roth prescription incorporates 12 degrees of palatal root torque (+12), while the MBT prescription includes 17 degrees of palatal root torque (+17).[1] The higher torque of the MBT prescription is advocated for based on the inefficiency of brackets in delivering torque, due to the known play in the system, as well as to prevent the retroclined appearance of the incisors.[19]

The amount of play in the system certainly influences the amount of torque expression as evidenced by previous studies. It is accepted that when using a full sized 0.019 x 0.025-in archwire in a 0.022-in bracket slot there will be approximately 10 degrees of play.[19] However, the measured value was found to be closer to 14 degrees[46], which would mean for a Roth prescription all of the built in torque may be lost.[35] Interestingly, Mittal et al[47] compared the torque achieved with MBT and Roth prescriptions clinically and found that the differing prescriptions did not amount to any clinically detectable differences in the final inclination of the upper incisors. This difference cannot be explained by the play alone, which demonstrates that, as stated earlier, torque expression is multi-factorial.

1.9.4 Tooth Morphology and Bracket Positioning

When the SWA was developed and used, it is based upon the assumption that the point of facial contour for each type of tooth is the same for all patients. However, it has been shown that there exists differences in the morphology of teeth, specifically the labial contour of the tooth, which can range from 2.6 to 6.4 degrees.[48] In addition, the labial contour of a given tooth varies from occlusal to gingival, such that the axial inclination produced when a full sized archwire is engaged in a bracket will vary depending on both the labial surface contour and the vertical height at which it is placed. More specifically, Van Loenen et al[49] reported that when bracket placement is varied within a clinically acceptable range of 2 mm along the surface of a maxillary central incisor, the difference in torque can be nearly 20 degrees. In general, a bracket that is placed more incisal will result in increased torque values when compared with the same bracket that is placed more gingival. If one is to attempt to eliminate this variable of torque expression, a fully customized bracket must be used.[50]

1.9.5 Bracket Slot Dimensions

As noted earlier, conventional bracket systems most commonly used are available in 0.018-in and 0.022-in slot dimensions. The preference for slot size varies around the world. A series of studies completed in the United States between 1986 and 2014 found the use of the 0.018-in slot has been declining, which is mirrored by an increase in the use of the 0.022-in slot, approaching use by 70% of respondents.[51] Generally, European orthodontists prefer the 0.018-in and practitioners in the United Kingdom and North America favour the 0.022-in systems. Advantages to both slot sizes have been proposed, such as overbite reduction and closure of extraction space using sliding mechanics may be more efficient with 0.022-in bracket slots, while the working archwire for 0.018-in bracket slots is presumed to deliver torque more effectively and earlier in treatment without additional wire bending.[52] These differences can be attributed to the amount of play present in each system when utilizing their corresponding full-sized archwire. The amount play between a 0.016 x 0.022-in working wire in the 0.018-in bracket slot is 7.8 degrees, whereas in the 0.022-in bracket system and its 0.019 x 0.025-in working wire the play amounts to 9.5 degrees.[53] The question remains as to whether this 1.7 degree difference amounts to significant differences in torque expression.

When comparing the effectiveness of the different bracket slot sizes, studies have shown that there is a slight decrease in treatment time when using the 0.018-in slot compared to the 0.022-in slot. This difference ranged from 2 to 4 months, depending on the study, of which both were found not to be clinically significant.[54,55] It was postulated that this small difference in treatment time could be accounted for by the difference in the bracket-wire play with the respective working archwire combinations, in that the full expression of the bracket prescription could be achieved earlier in the 0.018-in bracket slot.[54] With regards to incisor inclination, it has been shown that both bracket slots can produce statistically significant amounts of incisors proclination at the end of treatment. The 0.018-in brackets produced on average 1.3 degrees more proclination compared to the 0.022-in brackets, which was a small difference that is not considered clinically significant but was found to be statistically different.[56] When a 0.017 x 0.025-in archwire was utilized as the working archwire in a 0.018-in bracket slot and compared with a 0.019 x 0.025-in wire in a 0.022-in slot, and both combinations were torqued to 15 degrees, it was found that the 0.018-in system produced 14.3 Nmm of torque whereas the 0.022-in system was only capable of producing 9.3 Nmm of torque.[57] It has been proposed that bracket design alone (including slot size) has little effect on torque expression, whereas the combination of archwire and slot size are responsible for the expression of torque.[56]

1.9.6 Bracket Material

Orthodontic brackets are available in a range of different materials, including SS, titanium, plastic, ceramic, and composite plastics. As stated in Proffit[1], stainless steel brackets are still considered the standard option, and can be fabricated via metal injection molding (MIM), casting, or 3D printing. Most SS appliances are made via MIM, yet milling the slot of a cast bracket results in better precision of the bracket slot. Brackets produced via 3D printing have the potential to offer superior bracket slot precision. This technology is only being used to produce lingual SS appliances, for the time being. Titanium brackets can be offered as an alternative to SS brackets, a major indication of which is a sensitivity to the nickel content of SS. Titanium brackets also offer better bond reliability, as titanium is half as stiff as SS such that it can absorb more impact energy during function. This result of reduced load translates to less bond failures. A drawback of titanium brackets, due to titanium's inherently rough surface, is greater resistance to sliding mechanics if there is a large contact area between the bracket and the archwire.

In an attempt to produce more esthetic brackets, non-metallic options, such as plastic, ceramic, and composite plastic, have been produced. Plastic brackets have shown problems with staining and discolouration, poor strength and dimensional stability, and friction and binding between the bracket and metal archwires. To overcome some of these challenges plastic brackets can be fabricated with metal slots, yet it is still only recommended to use plastic brackets when complex tooth movements are not required. Ceramic brackets perform better than plastic in terms of durability and staining, and can be fabricated through custom molding while also being dimensionally stable, such that precise bracket slots and prescriptions can be incorporated. Issues with ceramic brackets include fracture of the bracket due to the brittle nature of ceramics, friction within the bracket slot, enamel wear due to patients occluding on the bracket, and enamel fracture during bracket removal.

The effect of bracket material on torque is derived from the differences in the stiffness and strength of the material used for the fabrication of the bracket slot. Soft and compliant materials have been shown to plastically deform during torque application, which thereby reduces the torque expression capacity of the bracket.[58] It has been shown that early generations of plastic brackets experienced deformation of their bracket slots when torquing forces were applied with rectangular archwires. The deformation experienced was shown to distort the bracket such than it would be incapable of transmitting an adequate amount of torquing force to a tooth.[59] Modifications to plastic brackets, such as metal slot reinforcement, have been shown to strengthen the brackets such that they are capable of producing adequate torque values comparable with metal brackets.[60] Ceramic brackets have been shown to produce high torquing moments and low torque loss, which was comparable with metal brackets. This was found to be due to the high modulus of elasticity of the ceramic material. Yet this increased stiffness was found to also result in more fractures of the brackets, particularly the tie wings, during testing, demonstrating the trade-offs that accompany the use of ceramic brackets.[41]

Plastic deformation of the bracket is not an exclusive phenomenon to plastic brackets, as it has been shown to occur in metal brackets as well. Depending on the bracket system investigated, it can be estimated that between 0.6 and 7.7 degrees of increased torque play can occur through increases in the slot taper, after a single application of torque.[61] However, it remains to be shown if this increase in torque play is clinically relevant, as Hixson et al[62] found that increases of 3 degrees resulted in no significant difference in torque expression. The interaction between the hardness of the bracket and wire material is also a factor that must be considered in the plastic deformation of the bracket. It has been shown that SS wires are 3.7 times harder than SS brackets, such that the bracket is the weak point in the bracket-wire system. This was shown to lead to both notching and widening of the bracket slot after loading.[63]

1.9.7 Ligation Method

As noted early, there are different methods of holding an archwire within the bracket slot. Conventional brackets utilize either steel or elastomeric ligatures to hold the archwire in the slot. Self-ligating brackets use a clip or door mechanism to retain the wire within the slot, depending on whether they are considered active or passive, respectively (Figure 6).

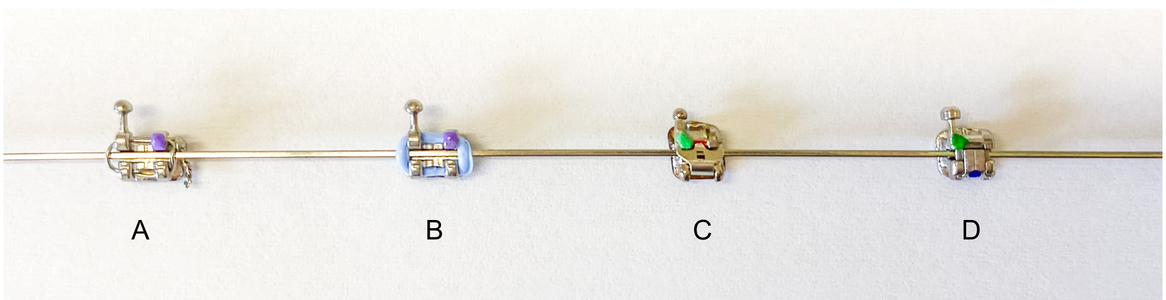


Figure 6: Examples of ligations methods, using a stainless steel ligature tie (A), elastomeric module (B), active self-ligation (C), and passive self-ligation (D)

Ligation of archwires to conventional brackets with a steel ligature wire, usually between 0.008 and 0.014-in in size[64], will rigidly hold the archwire in place and resist rotation. The SS ligature also has the capability to reinforce the bracket walls, to help resist the plastic deformation associated with torque expression that has been shown in several bracket systems. While the clinical relevance of the reduction of plastic deformation remains questionable, it has been shown that SS ligatures result in more immediate torque expression when compared with conventionally ligated (elastomer) and SL brackets.[65]

The torque moment with SS ligation was found to be 1.1 to 1.5 times greater than with elastomeric ligatures when undersized wires are considered. However, with full-sized wires, there was no difference found between the two ligation methods.[44]

Elastomeric ligatures are circular synthetic elastomers manufactured by injection molding or by being cut from previously processed elastomeric tubing. Beneficial characteristics of elastomeric ligatures include continuous gentle forces, long-lasting archwire seating, water sorption resistance, and shape memory. Disadvantages include binding during sliding mechanics and the possibility of archwires not being fully seated or engaged during torquing or rotational corrections. In an *in vitro* investigation by Taloumis et al[64] it was shown that elastomeric ligatures are affected by both moisture and heat, and exhibit rapid force loss of up to 68% in 24 hours. It was also shown that the elastomeric ligatures exhibit permanent deformation when stretched. They concluded that elastomeric ligatures are useful during initial leveling and alignment but were not effective at maintaining the archwire in the bracket slot during large rotational movements.

Self-ligating brackets were introduced as early as the 1930s, in an effort to save chair time associated with steel ligation methods. Since this time, the use of SL brackets has been promoted for time efficiency among other advantages such as low friction between the archwire and bracket slot for enhanced sliding mechanics, full archwire engagement, improved oral hygiene, and patient comfort.[15] The main difference between the two SL bracket types, active and passive, is the mechanism that closes the slot. Active self-ligating brackets (ASL) consist of a clip which exerts a pressure on the archwire. In contrast, passive self-ligating brackets (PSL) use a closing mechanism, much like a door, that transforms the open slot into a tube. Both bracket types claim advantages for their design: ASL brackets should enhance the control of tooth movement, as the pressure exerted on the archwire by the clip should produce better torque and rotational control; PSL brackets are promoted for their low friction mechanics.[31]

In comparing ASL and PSL brackets in terms of torque expression, Badawi et al[66] found that ASL brackets had better torque control. It was also found that ASL brackets engaged the archwires at earlier degrees of rotation, and were able to produce clinically effective torque (5 to 20 Nmm) at lower angles of torsion when compared to PSL bracket systems. The amount of bracket play was also considerably less for ASL brackets when compared to PSL systems, due to the active ligating mechanism. These findings are in contrast to Brauchlia et al[31] who found no significant difference between ASL and PSL systems, noting differences of torque expression of only 2 degrees which they claim would not be clinically significant. A systematic review on torque expression of SL brackets found only minor differences in torque expression between ASL and PSL brackets. This review also noted that conventionally ligated brackets produced higher torque expression than SL brackets.[24]

1.9.8 Degree of Wire Twist

Generally, as orthodontic archwires are engaged in bracket slots to produce torque, the torquing moments increase as the "twist angle" or degree of torque in the wire is increased.[31,34,42,44,65,66] However, some brackets have shown a decrease in torquing moments as torque angles increase[34,42], while other show a plateau at high torque angles.[66] Conversely, at low torque angles no torquing moments are generated in many bracket systems. This is due to the play in the system, such that the wire has not engaged the walls of the bracket slot and is unable to produce torque. Brauchli et al[31] have shown that in considering a 0.019 x 0.025-in archwire in a 0.022-in bracket slot, one would require between 20 and 25 degrees of twist to produce a torquing moment in the physiologic range of 5 to 20 Nmm for most bracket systems. Yet some systems tested required even higher degrees of twist, demonstrating the variability between bracket systems.

1.9.9 Direction of Wire Twist

Direction of wire twist has been an understudied factor of torque expression, yet there exist reasons as to why this may affect torquing moments. It has been shown that torquing moments are smaller when torque angles are decreasing compared to when they are increasing. It is thought that this occurs due to deformation of the archwire, the bracket slot, or both.[34] While one study has shown that the torque expression of some bracket systems is the same regardless of whether buccal root torque or palatal root torque are being expressed[31], others have shown that there is in fact a difference between the two depending on the bracket system tested.[67,68] In a recent thesis by Boogaards[68] it was shown that ASL brackets produced significantly lower torquing moments in a palatal root torque direction when compared with buccal root torque. Conventionally ligated and PSL brackets were not found to demonstrate differences between the two directions.

1.10 Methodologies for Studying Torque in Orthodontics

Studies on torque in orthodontic brackets have employed a variety of methods to measure torque expression.[23,24] One method commonly used is the orthodontic measurement and simulation system (OMSS), described by Drescher et al[69]. The OMSS consists of two force-moment sensors which are mounted on motor-driven positioning tables with full 3D mobility. All the mechanical components are housed withing a temperature-controlled chamber, and interfaced with a computer that records the resultant force-deflection curves. Typically, a dental model with brackets bonded to the teeth, into which an archwire can be placed, is used. The sensor of the OMSS replaces the tooth of interest in the arch, with the bracket bonded directly to the sensor. Another commonly used testing device was developed by Badawi et al[66] in 2008, which uses a multi-axis force/torque transducer capable of measuring forces and moments in three planes of space. It consists of a digital inclinometer to measure the degree of torsional rotation of the wire, a wire support substructure to hold the wire, a worm-gear to rotate the wire segment, and an alignment assembly which ensures proper alignment of the bracket and archwire, so that forces and moments other than torque are kept to zero.

Torque has also been studied using different styles of lathes and other novel apparatuses. In using lathes, inaccuracies can occur if the pulley does not fit tightly around the lathe, which will produce an axial force. To prevent this and consequent energy loss, a pulley that exerts a force couple could be used. It is also possible the wire may distort or twist within the lathe, leading to frictional torque loss.[23] Brauchli et al[31] bonded brackets to screw heads which were then mounted on the frame of a hexapod, while the archwire was fixed within a three-jaw drill chuck mounted on a 3D force/moment sensor. Young[67] developed a custom torque assembly fixed to an Instron materials testing machine to measure torquing moments generated with wire rotation. Most recently, Boogaards[68] created a custom, table-top torque assembly to evaluate torquing moments through the use of a small capacity load cell.

1.11 Gaps in the Literature

While there have been many studies which have evaluated torque expression in orthodontic brackets, there remains gaps in the literature when it comes to torque expression in SL bracket systems. There exists controversy as to whether the ligation method of a bracket does affect the torque expression, with some studies suggesting no difference between ASL and PSL, or no differences between SL and conventional brackets, while other studies have shown ligation method does affect the amount of torque expression. There is a particular lack of evidence when it comes to how the bracket-wire material combination affects torque expression. Most of the previous studies on torque expression in SL brackets have only utilized full-sized rectangular SS archwires. To our knowledge, only one study by Archambault et al[42] has investigated how different archwire materials affect torque expression. The findings of their study were presented as fractions of the maximum torque expression of a given bracket system, such that no available literature exists on expected torque moment values generated with different SL bracket-archwire material combinations.

Chapter 2

2 Purpose and Hypotheses

2.1 Purpose of the Current Investigation

The aim of the current investigation is to build upon the previous investigation completed in 2022 by Boogaards[68], and will compare torque expression using three materials of rectangular orthodontic archwire sizes in active, passive, and conventionally ligated 0.022-in orthodontic brackets, in both clockwise and counter-clockwise directions of wire rotation to determine if ligation method and archwire material influence how torque expression occurs. The result will enhance understanding of the influence of wire material, ligation method, and torque direction on torque expression in commonly available orthodontic appliances. It is anticipated this information will help guide orthodontists in choosing the appropriate bracket systems and wire materials to use for each patient, depending on the torquing needs of the case.

2.2 Hypotheses

1. Greater torquing moments will be generated by ASL bracket systems compared to PSL and conventionally ligated bracket systems, for all wire materials.

2. Clinically significant torque values will be more difficult to generate with TMA and NiTi, when compared to SS.

3. The direction of wire twist (buccal or palatal root torque) will not affect torquing moments generated with ASL, PSL, or conventionally ligated bracket systems, regardless of wire material.

4. Torsional stiffness values will be greatest for all SS wire-bracket combinations, followed by TMA, with NiTi wire-bracket combinations having the lowest values.

Chapter 3

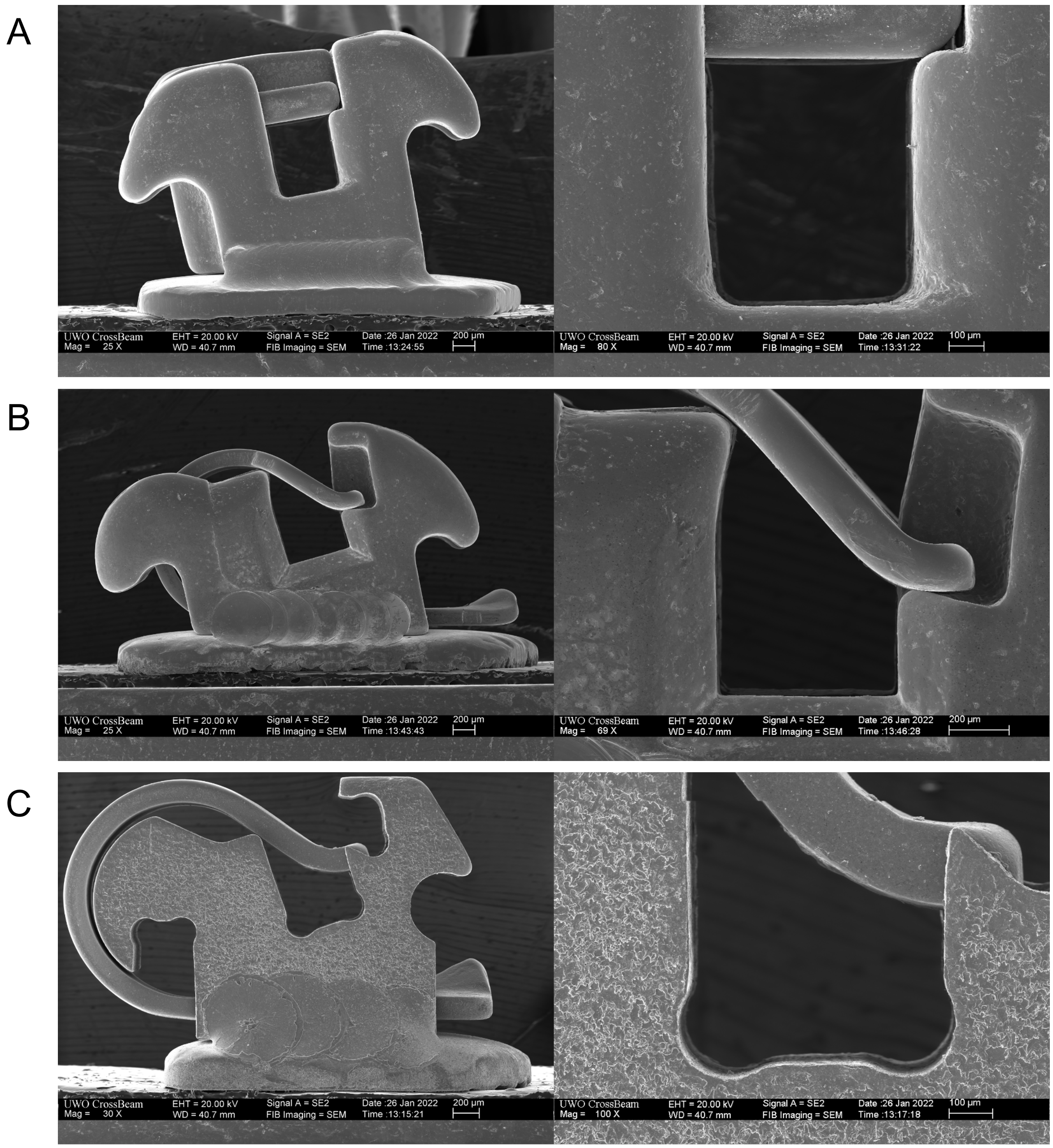
3 Materials and Methods

3.1 Brackets of Interest

Three 0.022-in ASL bracket systems, one 0.022-in PSL bracket system, and one 0.022-in conventional twin bracket system were tested in torque expression. More specifically, ASL systems tested included Empower®2 Metal Interactive, Speed System™, and 3M™ Victory Series™ ASL brackets. The PSL system studied was the Damon™ Q2 system. The conventional 3M™ Victory Series™ Twin bracket, ligated with an elastomeric module, was used as the control. These systems were chosen due to bracket popularity, availability, and use in the previous literature. All brackets tested were maxillary right central incisor brackets, with prescriptions that were the most commonly available for each system. Brackets were mounted for testing so that any incorporated torque prescriptions was zeroed, such that differences in torque prescription between systems did not influence results. A summary of the bracket systems tested are shown in Table 2. Scanning electron microscopy images of each bracket system are shown in Figure 7.

Table 2: Summary of features of bracket systems of interest

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	09224033 N
ASL	Empower®2 Metal Interactive	A-Emp	American Orthodontics (Sheboygan, WI, USA)	MBT	485-1117	P0019473
	Speed System™	A-Spd	Strite Industries (Cambridge, ON, Canada)	MBT	22UR1+1 7HR	011023 102522
	3M™ Victory Series™ ASL	A-Vic	3M™ Unitek™ Orthodontic Products (Monrovia, CA, USA)	Roth	025-302	NT7XP
Elastomeric	3M™ Victory Series™ Twin	C-Vic	3M™ Unitek™ Orthodontic Products (Monrovia, CA, USA)	MBT	017-876	NR5WA
	Elastomeric module	-	American Orthodontics (Sheboygan, WI, USA)	-	854-660	N37341



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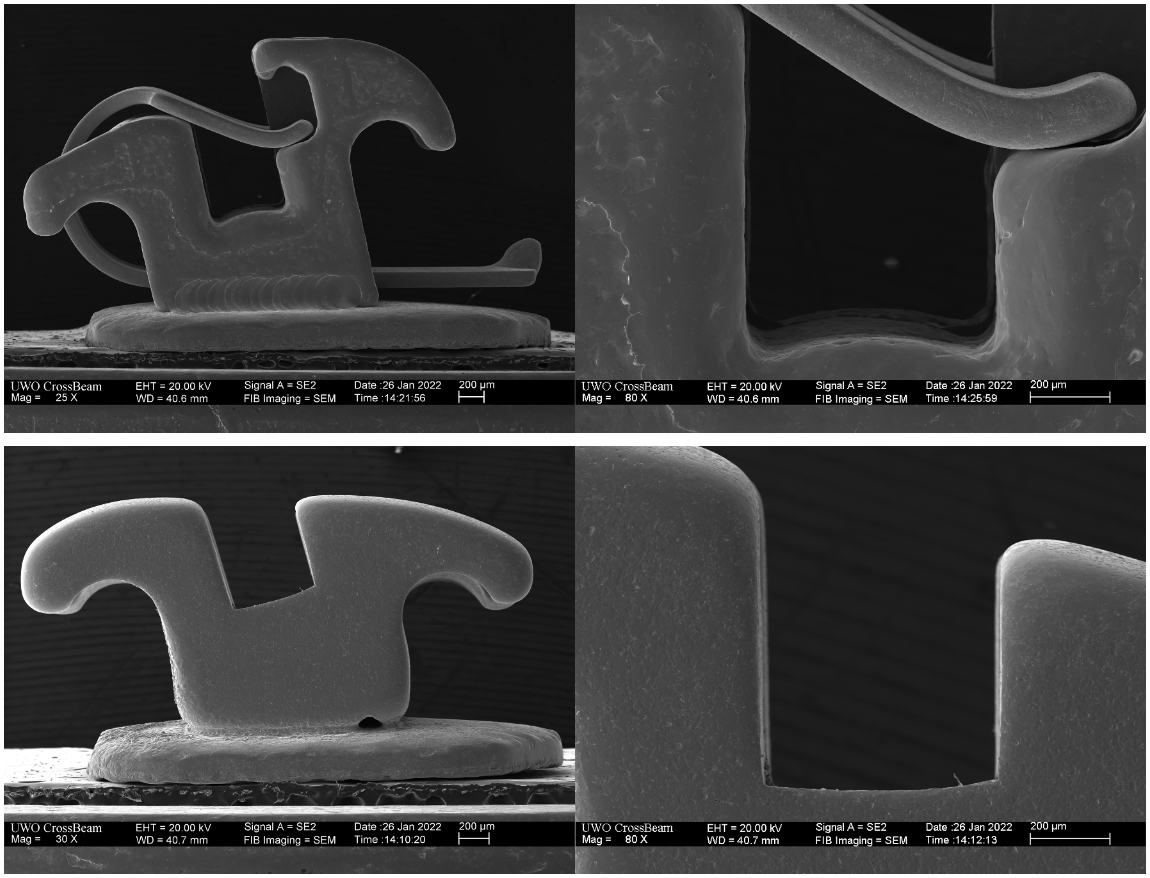

Figure 7: Scanning electron microscope images of each bracket system of interest, showing (A) P(Dmn), (B) A-Emp, (C) A-Spd, (D) A-Vic, and (E) C-Vic. Left images show the bracket as a whole with the gingival aspect oriented rightward, while the right images show a magnified view of the bracket slot. Measurements obtained from within the bracket slot are found in Appendices A-E.

3.2 Wires of Interest

Three wire materials were examined: NiTi, TMA and SS. Wires of these materials were selected due to their common use in clinical practice. Characteristics of the utilized wires are shown in Table 3. Wires utilized were each 0.019 x 0.025-in in size, based on the reported measurements from the manufacturer, G & H Orthodontics[70,71] (Franklin, IN, USA), in 7- and 14-in straight lengths. Force-deflection curves from the manufacturer can be found in Appendix F. Nominal dimension differences from manufacturer size designations has been reported in the literature, with actual wire sizes usually being less than the reported size.[32,72–74] The dimensions of the wires of interest utilized in this study were confirmed with SEM images, and found to be on average 0.001-in larger for both height and width (See Appendices G-I).

Table 3: Summary of features of investigated wire materials

Wire Material	Manufacturer	Item Number	Lot Number
Nickel titanium	G & H Orthodontics (Franklin, IN, USA)	SENT1925	912286
TitanMoly™ Titanium molybdenum		STB31925	1080321
Stainless steel		STSS1925	926631

3.3 Apparatus: Mounting Jig

A custom mounting jig as described by Boogaards[68], and adapted from previous studies by Young[67] and Greene et al[72], was utilized for the current investigation. It consisted of an aluminum base with a slot in the center, and connected two rectangular aluminum poles at either end, as shown in Figure 8. The slot in the base received a single hexagonal SS transfer pin, which was accurately positioned and secured with a screw to the jig. The aluminum poles each housed a clamp which would receive a 0.0215 x 0.025-in SS wire

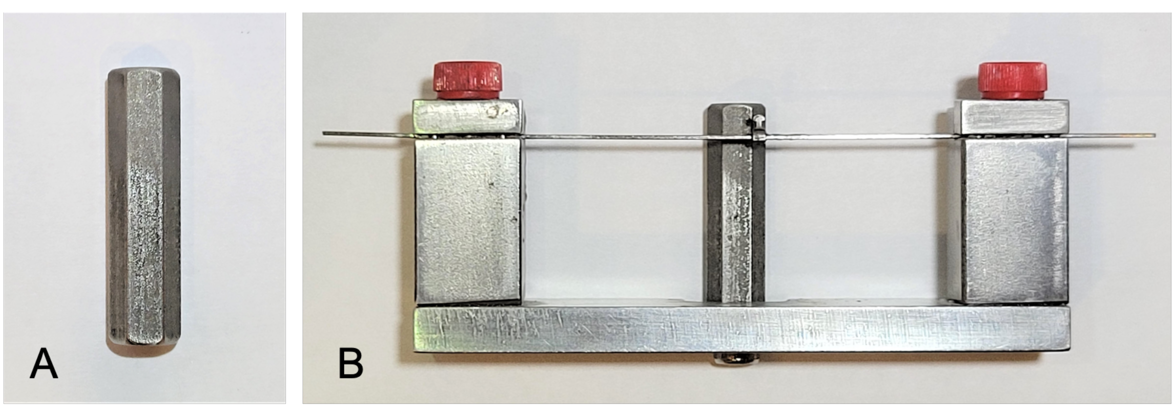

Figure 8: Prepared hexagonal transfer pin (A), and assembled mounting jig with a 0.0215 x 0.025-in SS wire and crimpable stop, and installed transfer pin (B)

used for bracket mounting. A wire of this dimension will have nearly completely filled the slot of the 0.022-in brackets, to eliminate the pre-programmed prescriptions and to allow each bracket to be mounted in precisely the same position. A crimpable stop was placed on the mounting wire, just offset of the midline of the jig, to ensure repeatable and precise positioning of the brackets onto the midpoint of the transfer pins. Brackets were secured to a transfer pin by first assembling the jig and transfer pin securely together, followed by inserting the mounting wire into the jig clamps, then ligating the bracket onto the wire and sliding it along the wire to the stop and bonding the bracket to the transfer pin.

3.3.1 Bracket Mounting Protocol

The protocol for bonding the tested brackets to the SS transfer pins began with micro-etching the surface of the transfer pins using 50 micron aluminum oxide (Item#15301, Lot#L0BWZ, Danville Materials, Carlsbad, CA, USA). Assure® Plus All Surface Light Cure Bonding Primer (Item#PLUS, Lot#215151, Reliance® Orthodontic Products, Itasca, IL, USA) was applied and air thinned on the transfer pin surface, after which GoTo™ light cure adhesive (Item#GTP, Lot#214169, Reliance® Orthodontic Products, Itasca, IL, USA) was applied to the bracket base and transfer pin. Brackets were aligned relative to the transfer pins using the mounting jig, where it was verified that the occlusal wall of the bracket slot was parallel with the mounting wire, to ensure repeatable positioning. Excess adhesive was removed and the remainder was light cured. All materials were handled with gloves during the bonding procedure, to ensure no contaminants were introduced.

For each wire material tested, 10 brackets of each system were mounted for testing with the gingival aspect of the bracket oriented upwards, and another 10 were mounted with the gingival aspect of the bracketed oriented down. This allowed testing to simulate both palatal root torque (movement of the root towards the patient’s tongue or palate) and buccal root torque (movement of the root towards the patient’s lip or cheek), respectively.

3.4 Apparatus: Custom Torque Assembly

A custom, table-top torque assembly was fabricated to evaluate torquing moments generated by various bracket-wire combinations for the study completed by Boogaards[68]. Simply put, the device involved controlled rotation of individual brackets with respect to a fixed segment of the chosen wire of interest. The device was fabricated such that each wire of interest could be centered in the slot of the bracket while also remaining parallel to the axis of rotation of the stepper motor and torque measuring component. The assembly was fabricated from a combination of aluminum and 3D printed plastic components, which underwent element analysis to demonstrate that stress and deformation associated with a maximum expected torque of 250 Nmm would not significantly impact measured torque or result in long-term fatigue or distortion. This same device was modified slightly for use in the current investigation, mainly in that a limit-switch was added and the programing was modified slightly.

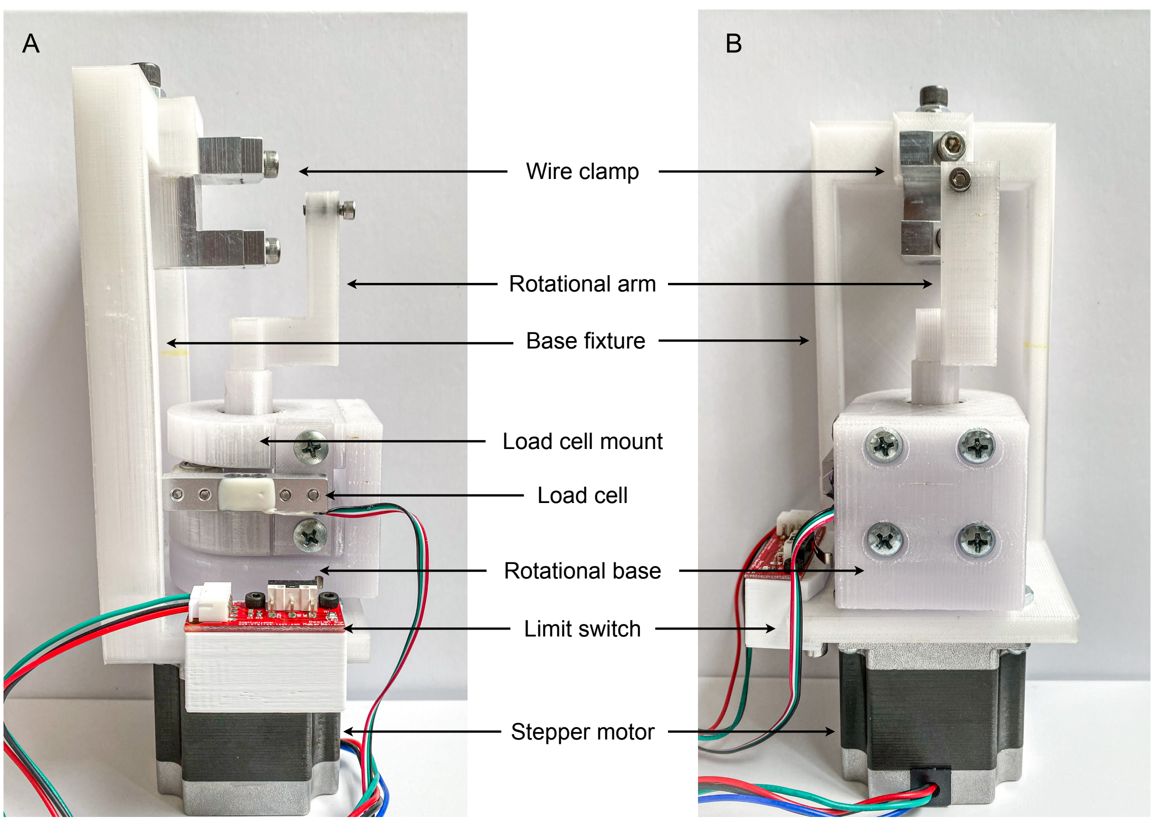

Figure 9: Custom torque assembly viewed from side (A) and front (B) perspectives. Labels indicate the main components of the assembly

More specifically, the device is composed of a stepper motor; the torque-measuring fixture, which included a load cell and a load cell mount fixed on a rotational base; a limit switch; a rotational arm, on which the transfer pin with mounted brackets could be added; and a base fixture with custom wire clamps that firmly held the wire of interest in position relative to the other elements (Figure 9).

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 to hold the wire of interest stationary throughout testing. The dimensions of the wire clamp were fabricated to be 0.018x0.024 inches, 0.001 inch 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.[32,72–74] The clamp was fastened to the base fixture using a dowel and screw. The span of wire between the clamps was precisely 15 mm to replicate the span of wire used to torque an average upper right maxillary central incisor in clinical practice. Aluminum 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. The limit switch was mounted to the side of the base fixture, using bolts. The limit switch was positioned to allow for precise positioning of the rotational base in a repeatable zero position before each test was conducted.

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 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 10, with transparency added to highlight the innerworkings of this component.

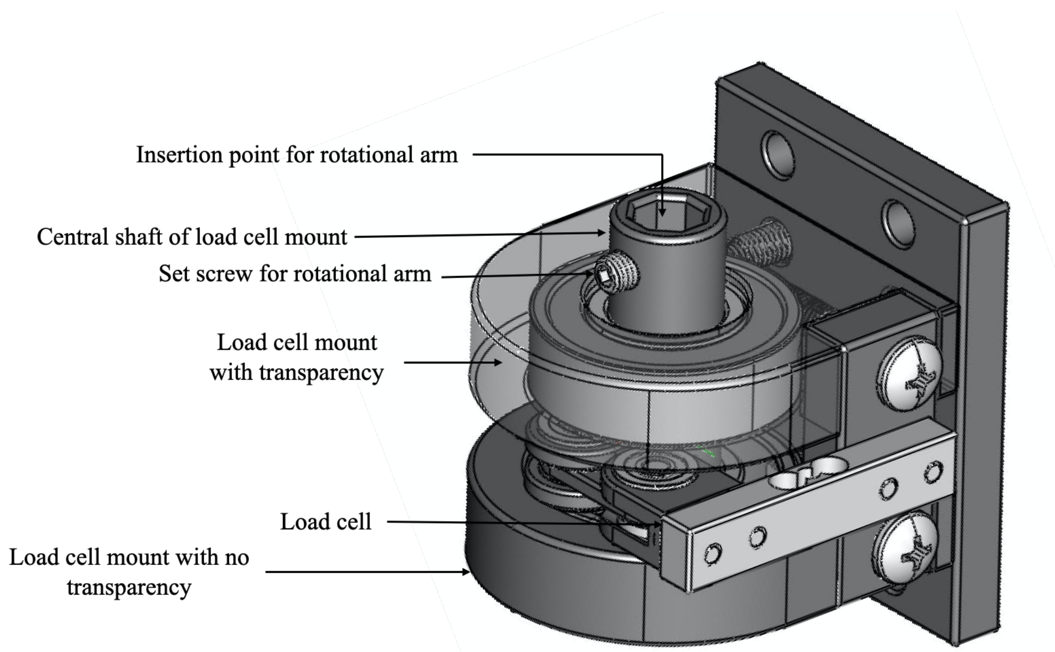

Figure 10: Isometric CAD view of the load cell mount with transparency to highlight the innerworkings of the torque measuring fixture

The device was controlled using a microprocessor (Arduino® Uno board) and Arduino® programming via a personal computer. Within the Arduino® programming, the target position in degrees and speed of rotation in degrees per second could be specified, in addition to a tare feature which was used to zero the torque measurements prior to each test. 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 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. The torque assembly was tared to zero using the Arduino® programming, and then the bracket was secured onto the wire. The device was then homed to its zero position, after which a load cell reading was measured and recorded. This "no-load" torque value was 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 Arduino® programming. Once the command was input, the stepper motor subsequently initiated rotation, resulting in rotation of the bracket of interest around the wire of interest. Recorded torque values were sampled from the load cell at increments of 0.11 degrees. For each test, the rotational arm was brought to 15 degrees beyond the zero position (denoted negative) in a clockwise direction (as viewed from above the custom torque assembly), as an additional check that the wire was centered in the bracket slot. From this point the test began, with rotation at a rate of 1 degree per second. Each test brought the bracket from the -15 degree position through zero and to its maximum +45 degree position via counterclockwise rotation, which was known as the loading data. The bracket was then subsequently brought back along the same path in a clockwise direction from +45 degrees to -15 degrees, known as the unloading data. The resulting data was automatically populated to an Arduino® output window specifying the current position in degrees, and torquing moments in Nmm.

For each wire material 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. 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 A-Emp(B) represented an Empower ASL bracket undergoing a simulation of buccal root torque). A fresh bracket and wire segment was used for each replicate, and in addition, for conventionally ligated systems, a fresh elastomeric ligature was utilized for each replicate. Tests were all performed at room temperature.

3.6 Calibration of the Custom Torque Assembly

Before its first use, the load cell was calibrated utilizing multiple known weights (each 4.69 g). The weights were 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, confirmation of calibration was assessed using known weights of 200 and 500 g suspended from the rotational arm of the apparatus, of which the torque readings were compared to readings obtained immediately following initial calibration. Torquing moments generated by the known weights were recorded before and after each testing group and compared to initial readings of 54 Nmm and 137 Nmm for the 200 g and 500 g weights respectively. If drift in calibration was detected, a correction factor could be applied to the resultant data if needed. However, no correction factor was needed as the measurements remained consistent throughout testing. The 200 g weights produced torque measurements with a range of 2.7 Nmm and 500 g weights produced measurements with a range of 3.6 Nmm. These ranges were deemed adequately precise given clinically significant torquing moments measure a minimum of 5 Nmm.

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. 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, 3MTM 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. Descriptive statistics, including mean torquing moments and standard deviations, were calculated for each bracket-wire combination at the nearest approximation to 15 degree intervals in both the loading (15, 30, and 45 degrees) and unloading (30' and 15' degrees) directions. Normal distribution of the data was confirmed using statistical software (IBM® SPSS® Statistics 27.0; SPSS, Inc., Chicago, IL), such that torquing moment data at these intervals could be evaluated 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 material) on mean torquing moments generated at each 15 degree increment. 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 ($F(18,270)=22.57$, $P<0.001$); 30 degrees ($F(18,270)=51.06$, $P<0.001$); 45 degrees ($F(18,270)=58.16$, $P<0.001$); 30' degrees ($F(18,270=44.69$, $P<0.001$); 15' degrees($F(18,270)=16.77$, $P<0.001$)]. Therefore, One-Way ANOVAs with Bonferroni correction for multiple comparisons were conducted to uncover simple main effects of both bracket system and wire material.

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 material combination in the loading direction using computer coding. 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 11). Mean engagement angles were evaluated using Two-Way ANOVA with Bonferroni correction for multiple comparisons to assess the effect of both independent variables (bracket system and wire material) on engagement angles. 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 material ($F(18,270)=7.37$, $P<0.001$). Again, One-Way ANOVAs with Bonferroni correction for multiple comparisons were conducted to uncover simple main effects.

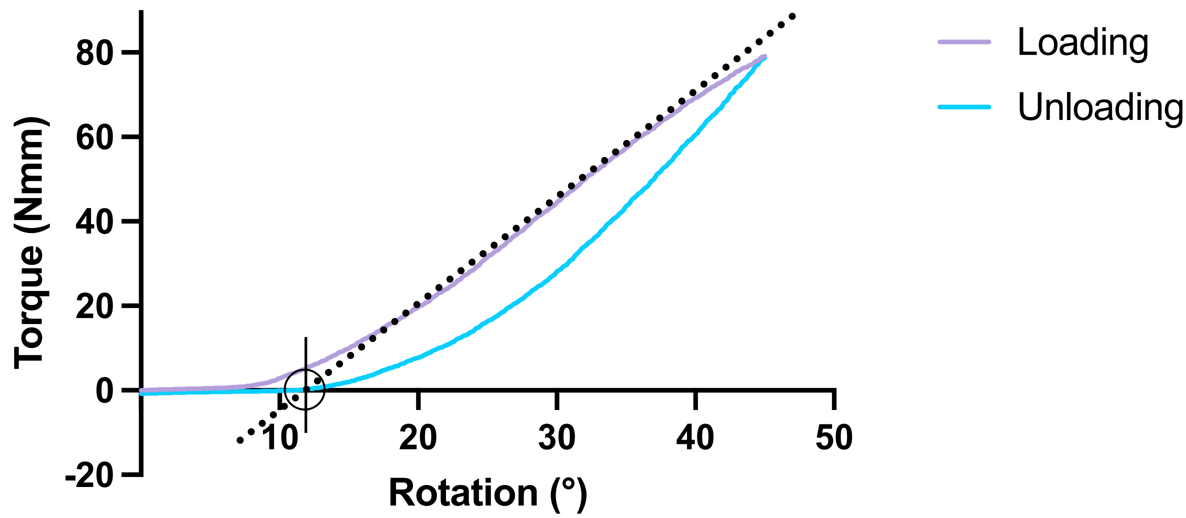

Figure 11: Engagement angle was determined by computer code, in which a best fit line was determined for the linear portion of the torque-rotation loading curve. The angle at which this line passed through zero was assigned as the engagement angle

3.7.3 Torsional Stiffness

To evaluate the torsional stiffness of each bracket-wire combination, a mean slope and standard deviation was determined for each bracket system-wire material combination. The slope was measured from the unloading curve, at its theoretical maximum between 40' and 45' degrees (Figure 12). Mean slopes were evaluated using Two-Way ANOVA with Bonferroni correction for multiple comparisons to assess the effect of both independent variables (bracket system and wire material) on the slope values. 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 material ($F(18,270)=30.18$, $P<0.001$). One-Way ANOVAs with Bonferroni correction for multiple comparisons were again conducted to uncover simple main effects.

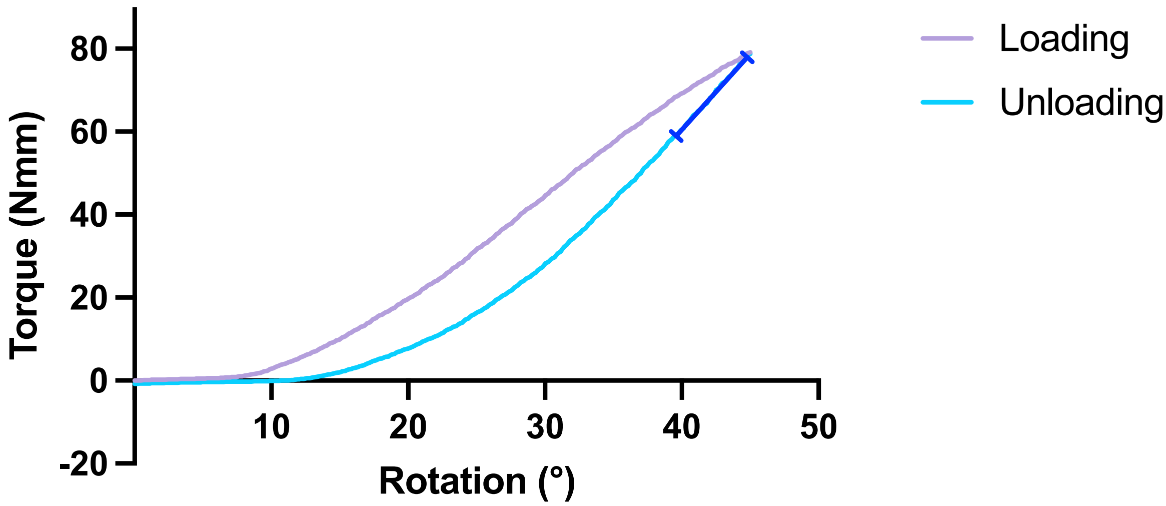

Figure 12: Torsional stiffness as determined by the slope of the unloading curve between 40' and 45' degrees

3.7.4 Hysteresis

To evaluate energy loss that occurred in the system over the course of each trial, a mean hysteresis value and standard deviation was determined for each bracket system-wire combination. Differences in values of torque at 30 degrees of rotation on both the loading and unloading curves were used to represent the change in the system (Figure 13). Mean hysteresis values were evaluated using Two-Way ANOVA with Bonferroni correction for multiple comparisons to assess the effect of both independent variables (bracket system and wire material) on the hysteresis values. 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 material ($F(18,270)=5.79$, $P<0.001$). One-Way ANOVAs with Bonferroni correction for multiple comparisons were conducted to uncover simple main effects

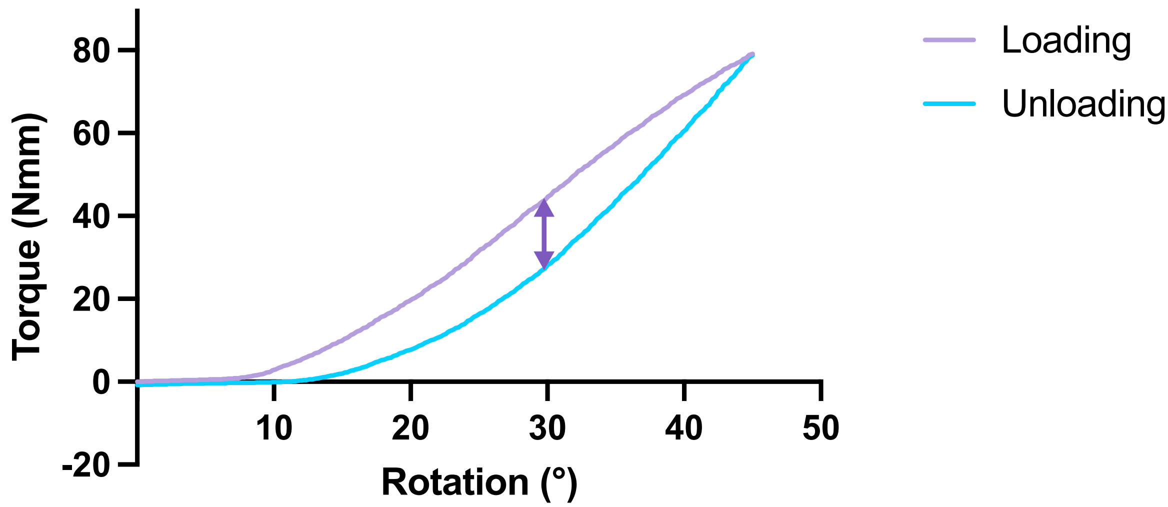

Figure 13: Hysteresis as determined by noting the difference in torquing moments between loading and unloading curves at 30 degrees of rotation

Chapter 4

4 Results

4.1 Torquing Moments

Mean torquing moments in Nmm determined for every 0.11 degree increment of rotation from 0 through +45 degrees (loading curves) for each bracket-wire combination in both directions of rotation (buccal and palatal root torque experiments) are plotted in graphical format as torque-rotation curves in Figures 14-16. Mean torquing moments for each bracket-wire combination at 3 degree intervals of rotation are included in Appendices J-L.

Generally, 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 TMA and SS wires, the relationship between torque and degree of rotation was generally linear after engagement of the wire within the slot, with the exception of the A-Spd/SS group in the buccal root torque direction. For NiTi wires however, beyond approximately 30 degrees of rotation, the linear trend between torque and degree of rotation was not maintained, and the curve began to reduce in slope at high degrees of rotation.

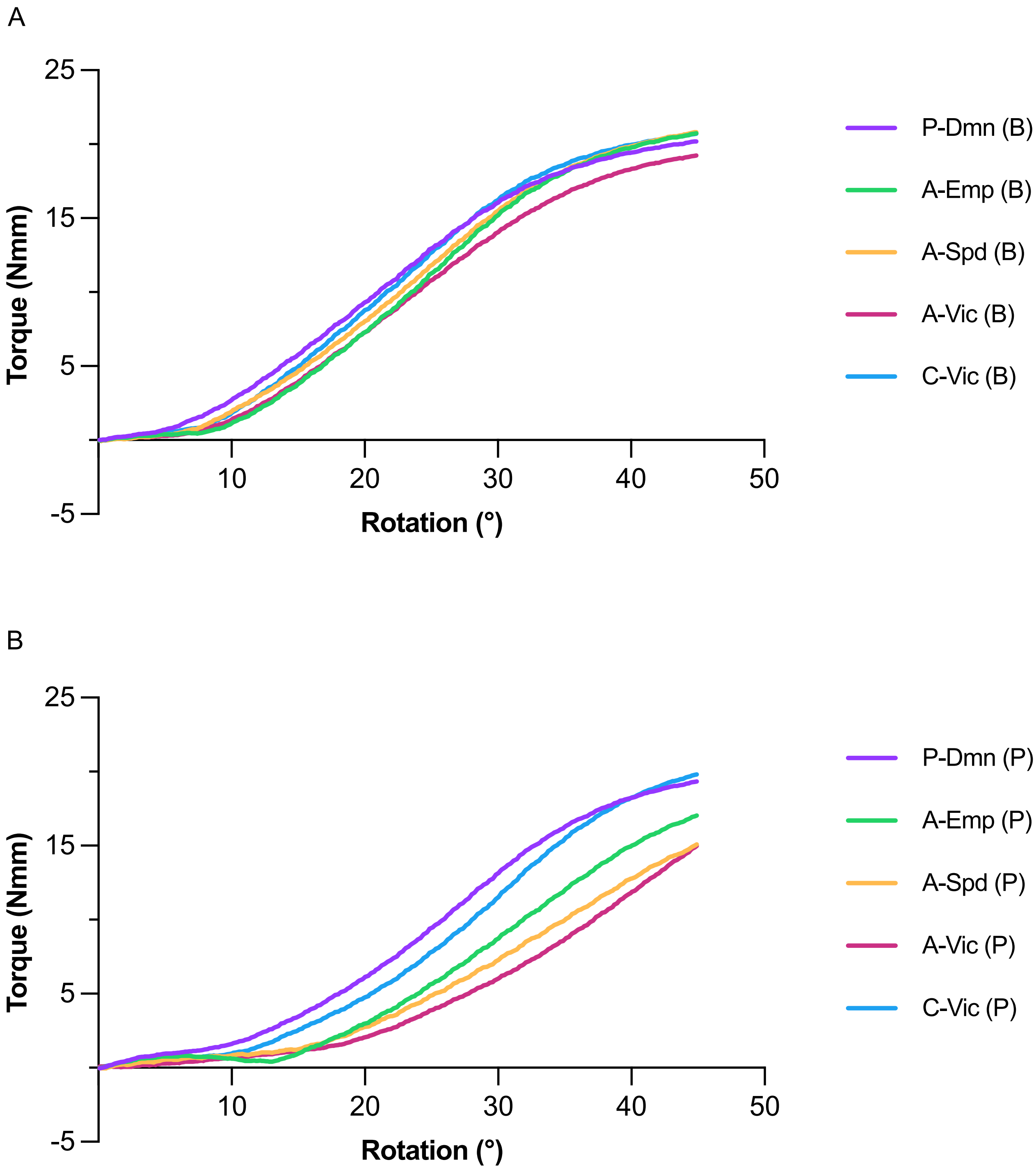

Figure 14: Mean torquing moments measured for each bracket system and NiTi wires, demonstrating both buccal (A) and palatal (B) root torque

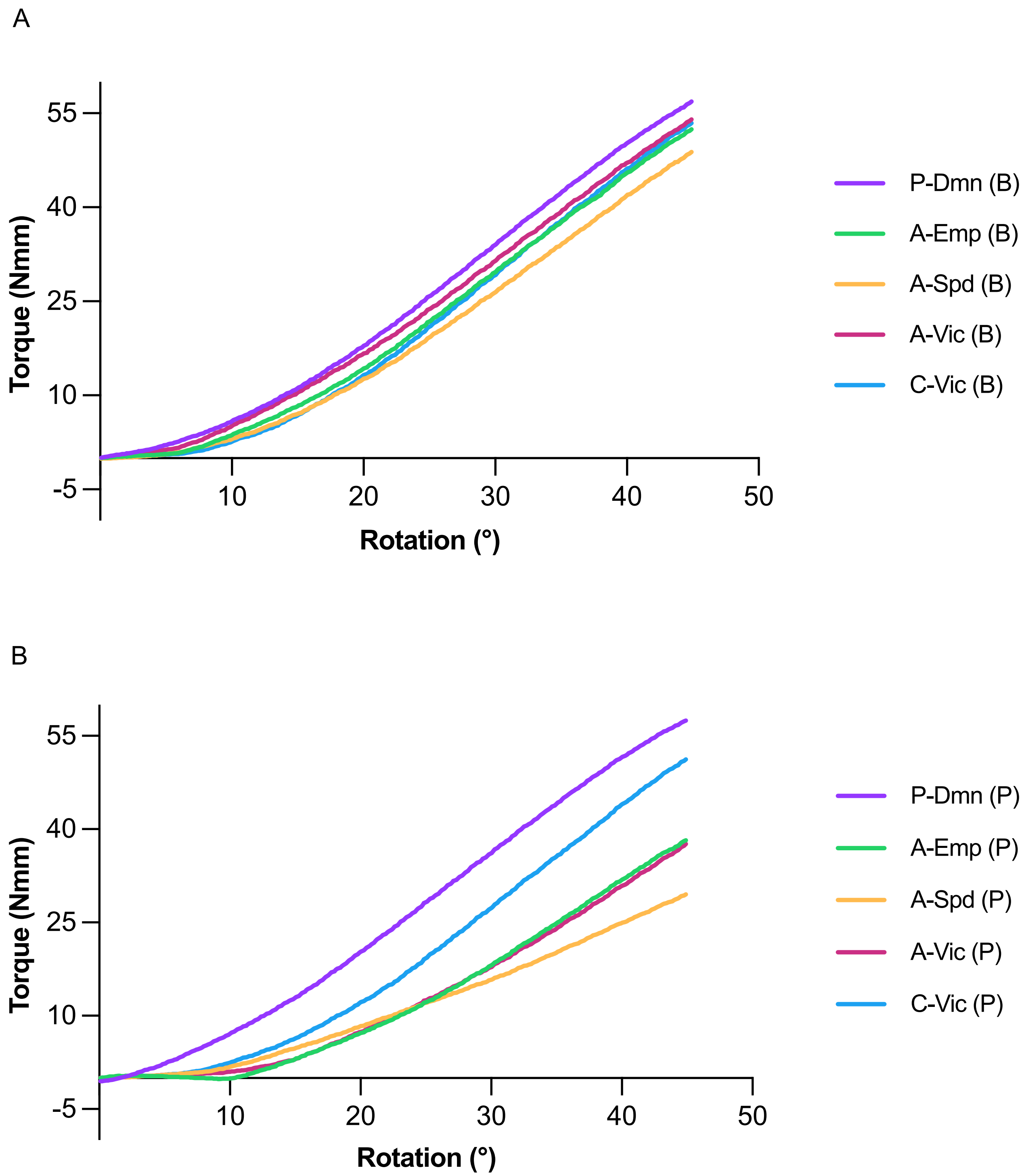

Figure 15: Mean torquing moments measured for each bracket system and TMA wires, demonstrating both buccal (A) and palatal (B) root torque

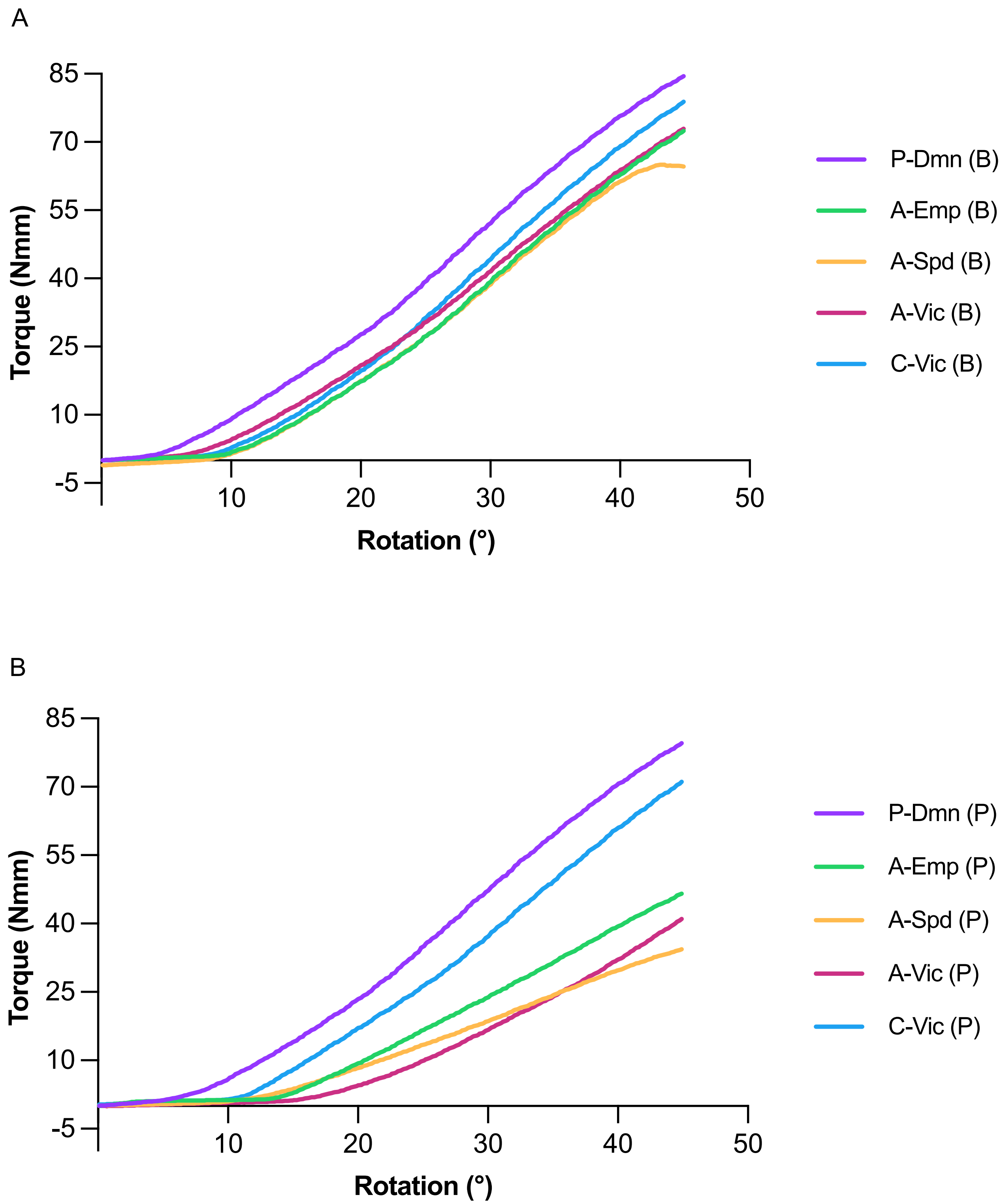

Figure 16: Mean torquing moments measured for each bracket system and SS wires, demonstrating both buccal (A) and palatal (B) root torque

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 material showed a statistically significant difference between bracket groups for all comparisons ($P<0.001$). Associated means (±SD) with significance results are found in both Table 4 and Figures 17-21, and associated P-values are included in Table 5.

In NiTi wires, for all degrees of rotation, mean torquing moments generated are not significantly different between bracket systems in the buccal root torque direction. However, P-Dmn and C-Vic generated significantly larger torquing moments than the ASL bracket systems in palatal root torque directions for all degrees of rotation ($P <0.05$).

With TMA wires, the same pattern was observed in which P-Dmn and C-Vic generated significantly larger torquing moments when compared to ASL brackets in the palatal root torque dimension, for all degrees of rotation ($P<0.01$). In the buccal root torque direction, at 15 degrees of rotation P-Dmn and A-Vic generated significantly larger torquing moments ($P<0.05$), but at high degrees of rotation generally no significant differences were noted between bracket systems.

In SS wires, at 15 and 30 degrees of rotation P-Dmn generated significantly larger torquing moments when compared with all bracket systems, in both directions ($P<0.001$). At 45 degrees, both P-Dmn and C-Vic generated significantly larger torquing moments than the ASL systems in palatal root torque direction ($P<0.001$). In the buccal root torque direction, P-Dmn and C-Vic torquing moments were larger than the majority of the ASL bracket systems.

Table 4: Mean torquing moments in Nmm (±SD) generated for each bracket system-wire combination for every 15 degree increment of rotation for both loading and unloading, where unloading values are represented by prime (′) values. Non-significant differences between different bracket groups within a wire material-rotational increment at P>0.05 are denoted by shared alphabetical letters within each row

		Bracket Group									
Wire Material	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)
NiTi	15	5.7(1.4)^{e}	3.4(0.7)^{bc}	3.8(0.8)^{bcd}	0.9(0.4)^{a}	4.6(1.3)^{cde}	1.2(0.2)^{a}	3.9(1.2)^{cd}	1.1(0.3)^{a}	4.9(0.8)^{de}	2.5(0.2)^{b}
	30	16.1(1.1)^{f}	13.2(1.4)^{cd}	15.3(0.8)^{ef}	8.8(1.2)^{b}	15.5(1.3)^{ef}	7.3(0.9)^{ab}	14.1(1.4)^{de}	6.1(1.0)^{a}	16.3(0.7)^{f}	11.5(0.9)^{c}
	45	20.2(1.0)^{cd}	19.3(0.8)^{c}	20.8(0.4)^{d}	17.0(1.4)^{b}	20.8(0.7)^{d}	15.1(0.8)^{a}	19.3(0.8)^{c}	15.1(1.1)^{a}	20.7(0.4)^{d}	19.8(0.7)^{cd}
	30'	10.7(0.9)^{de}	10.0(0.7)^{cd}	10.7(0.3)^{de}	6.9(1.0)^{b}	11.2(0.7)^{e}	5.2(0.6)^{a}	10.2(0.6)^{cd}	5.0(0.8)^{a}	11.2(0.3)^{e}	9.5(0.7)^{c}
	15'	4.3(1.1)^{e}	2.8(0.5)^{bc}	2.4(0.8)^{b}	-0.1(0.3)^{a}	3.6(1.1)^{cd}	0.9(0.4)^{a}	2.7(1.0)^{bc}	0.7(0.2)^{a}	4.0(0.7)^{e}	2.2(0.3)^{b}
TMA	15	11.1(2.6)^{e}	12.8(2.9)^{e}	8.2(0.8)^{cd}	3.0(1.0)^{a}	7.0(1.7)^{bc}	4.8(1.6)^{ab}	10.4(2.3)^{de}	3.0(1.4)^{a}	6.9(1.4)^{bc}	6.3(0.9)^{bc}
	30	34.1(4.2)^{de}	36.2(3.2)^{e}	30.0(1.4)^{bc}	18.3(1.1)^{a}	26.7(4.6)^{b}	15.9(1.7)^{a}	31.5(3.4)^{cd}	17.9(2.5)^{a}	29.5(1.6)^{bc}	27.6(1.8)^{bc}
	45	56.9(3.4)^{e}	57.5(2.3)^{e}	52.7(1.4)^{cd}	38.5(0.9)^{b}	49.1(4.9)^{c}	29.5(2.3)^{a}	54.0(2.3)^{de}	37.6(3.3)^{b}	53.7(1.2)^{de}	51.5(2.3)^{cd}
	30'	30.3(2.8)^{ef}	31.3(2.2)^{f}	26.5(1.0)^{cd}	14.9(0.8)^{b}	24.5(4.2)^{c}	10.2(1.4)^{a}	27.9(2.7)^{de}	15.0(2.3)^{b}	26.5(0.8)^{cd}	24.5(1.7)^{c}
	15'	8.0(1.8)^{de}	8.8(1.8)^{e}	5.4(0.9)^{bc}	0.7(0.6)^{a}	5.0(1.6)^{b}	1.4(0.5)^{a}	7.1(1.8)^{cd}	1.6(0.5)^{a}	4.7(0.8)^{b}	4.2(0.7)^{b}
SS	15	18.0(1.7)^{e}	14.1(2.7)^{d}	8.2(2.3)^{b}	2.8(0.6)^{a}	8.1(2.5)^{b}	3.7(1.2)^{a}	11.9(2.3)^{cd}	1.2(0.4)^{a}	9.9(3.5)^{bc}	7.9(1.3)^{b}
	30	52.2(2.3)^{f}	47.7(4.7)^{e}	39.3(3.0)^{c}	23.9(1.0)^{b}	38.8(2.9)^{c}	18.8(3.1)^{a}	41.6(3.8)^{cd}	16.7(2.6)^{a}	44.7(3.7)^{de}	37.2(3.3)^{c}
	45	84.8(1.9)^{f}	79.9(3.8)^{ef}	72.4(2.3)^{cde}	46.6(1.3)^{b}	64.6(14.2)^{c}	34.3(4.4)^{a}	73.0(4.5)^{de}	41.0(3.5)^{ab}	79.1(2.8)^{def}	71.1(4.8)^{cd}
	30'	35.5(1.6)^{e}	32.0(3.0)^{de}	24.2(1.5)^{bc}	8.9(0.9)^{a}	21.3(9.6)^{b}	6.5(1.0)^{a}	26.5(2.8)^{c}	8.1(1.3)^{a}	28.2(2.0)^{cd}	23.6(2.8)^{bc}
	15'	3.6(1.2)^{d}	3.4(1.0)^{d}	0.40(0.6)^{a}	0.1(0.8)^{a}	1.0(1.2)^{ab}	1.0(0.2)^{ab}	2.6(0.8)^{cd}	0.8(0.2)^{a}	2.1(1.0)^{bc}	1.2(0.5)^{ab}

Table 5: Significance results comparing mean torquing moments between bracket systems within a wire material-rotational increment

Wire Material	Degree	Significance Results
NiTi	15	F(9, 90) = 39.36, P<0.001
	30	F(9, 90) = 120.72, P<0.001
	45	F(9, 90) = 71.37, P<0.001
	30′	F(9, 90) = 128.01, P<0.001
	15′	F(9, 90) = 41.31, P<0.001
TMA	15	F(9, 90) = 34.21, P<0.001
	30	F(9, 90) = 64.95, P<0.001
	45	F(9, 90) = 126.62, P<0.001
	30′	F(9, 90) = 105.03, P<0.001
	15′	F(9, 90) = 53.26, P<0.001
SS	15	F(9, 90) = 65.04, P<0.001
	30	F(9, 90) = 149.09, P<0.001
	45	F(9, 90) = 94.75, P<0.001
	30′	F(9, 90) = 84.48, P<0.001
	15′	F(9, 90) = 22.81, P<0.001

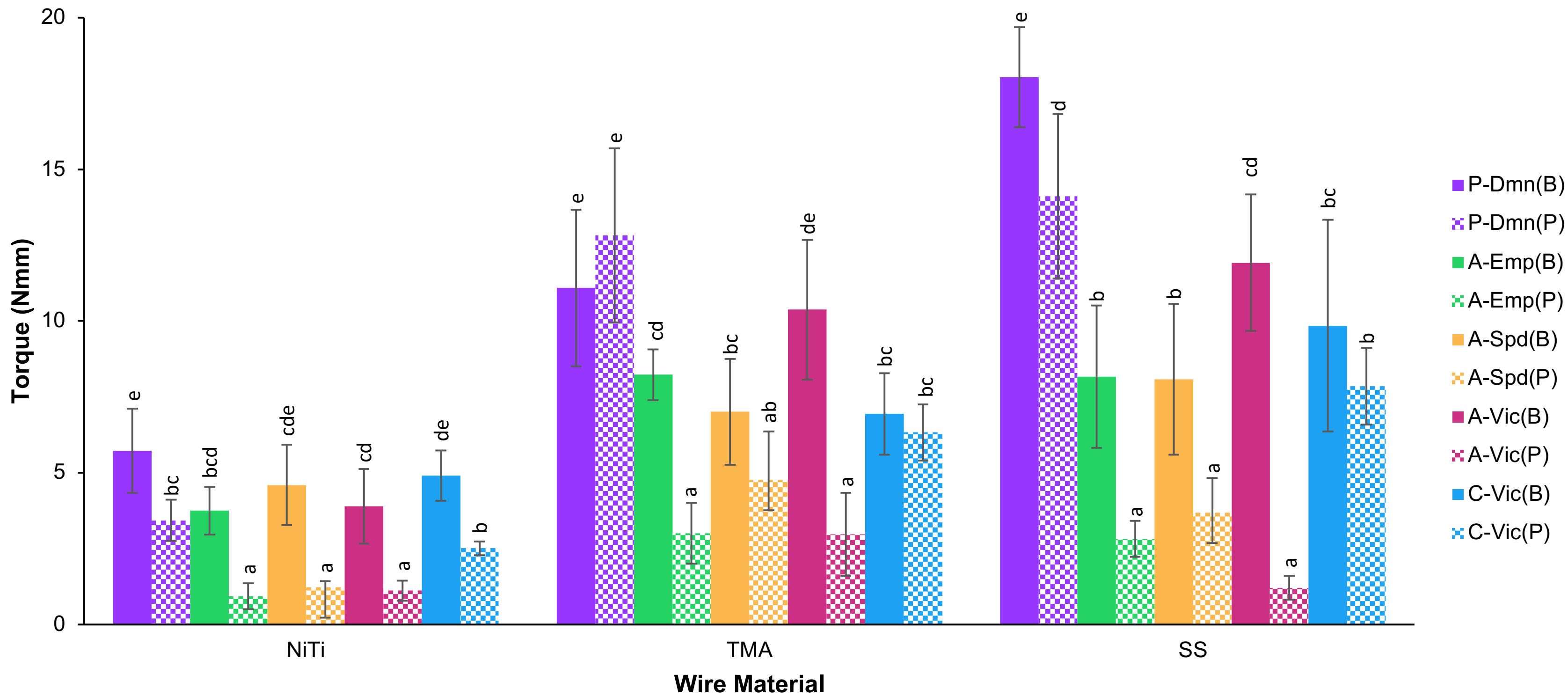

Figure 17: Mean torquing moments in Nmm (±SD) measured at 15 degrees with different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

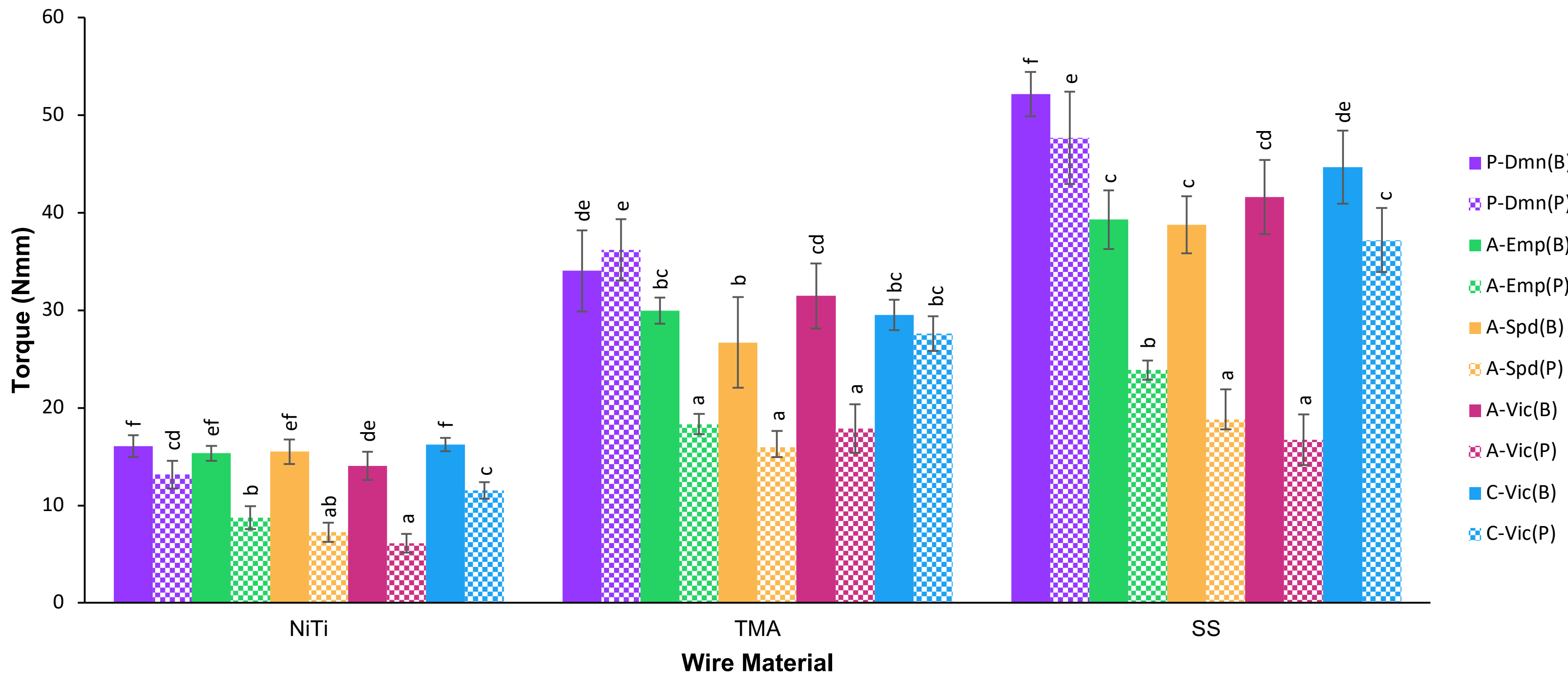

Figure 18: Mean torquing moments in Nmm (±SD) measured at 30 degrees with different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

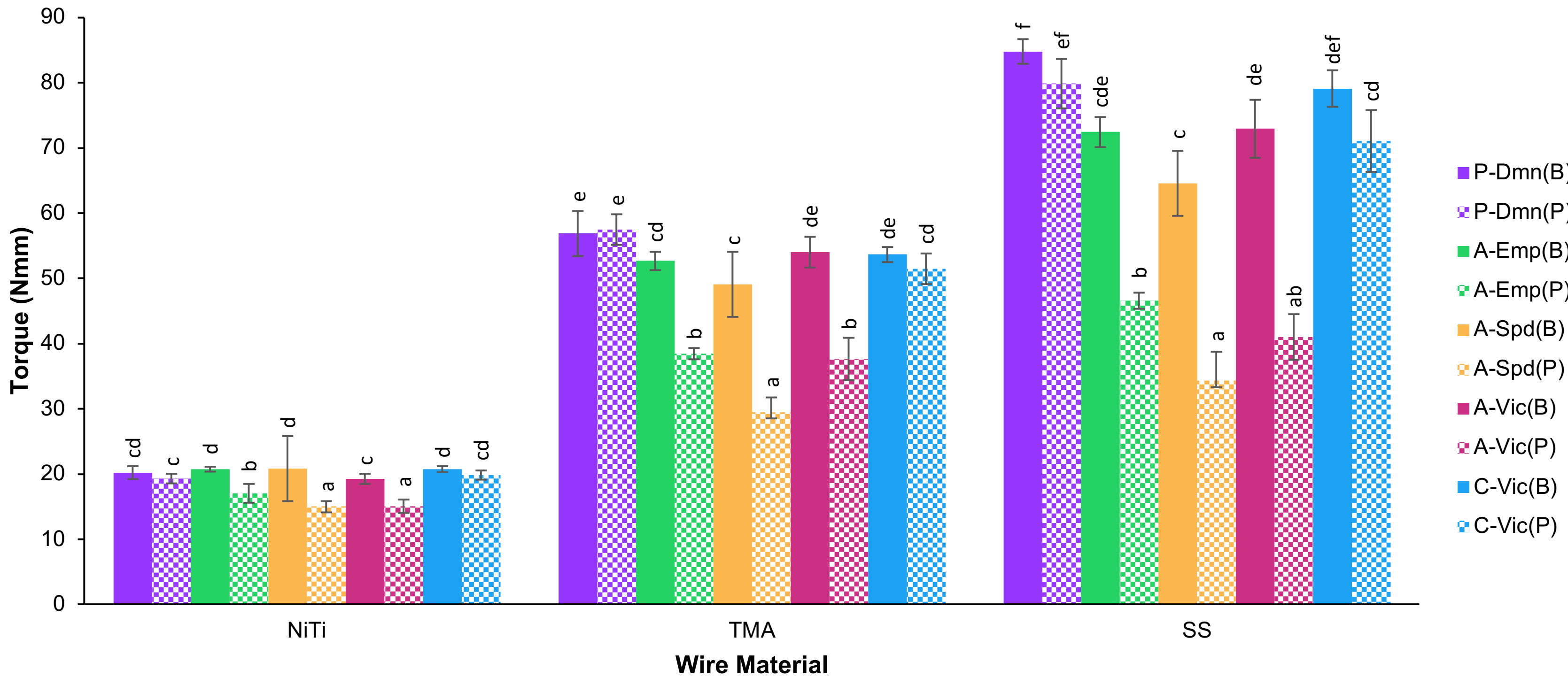

Figure 19: Mean torquing moments in Nmm (±SD) measured at 45 degrees with different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

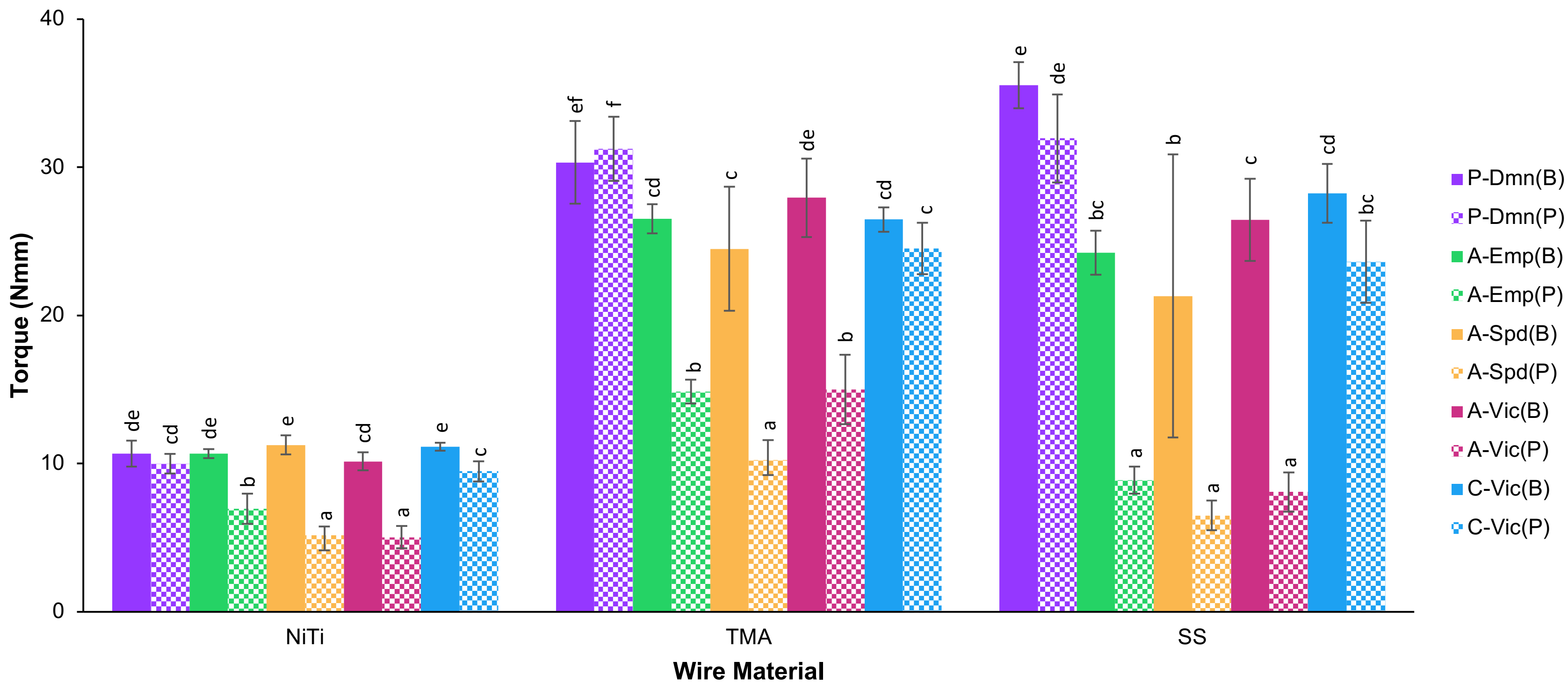

Figure 20: Mean torquing moments in Nmm (±SD) measured at 30' degrees with different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

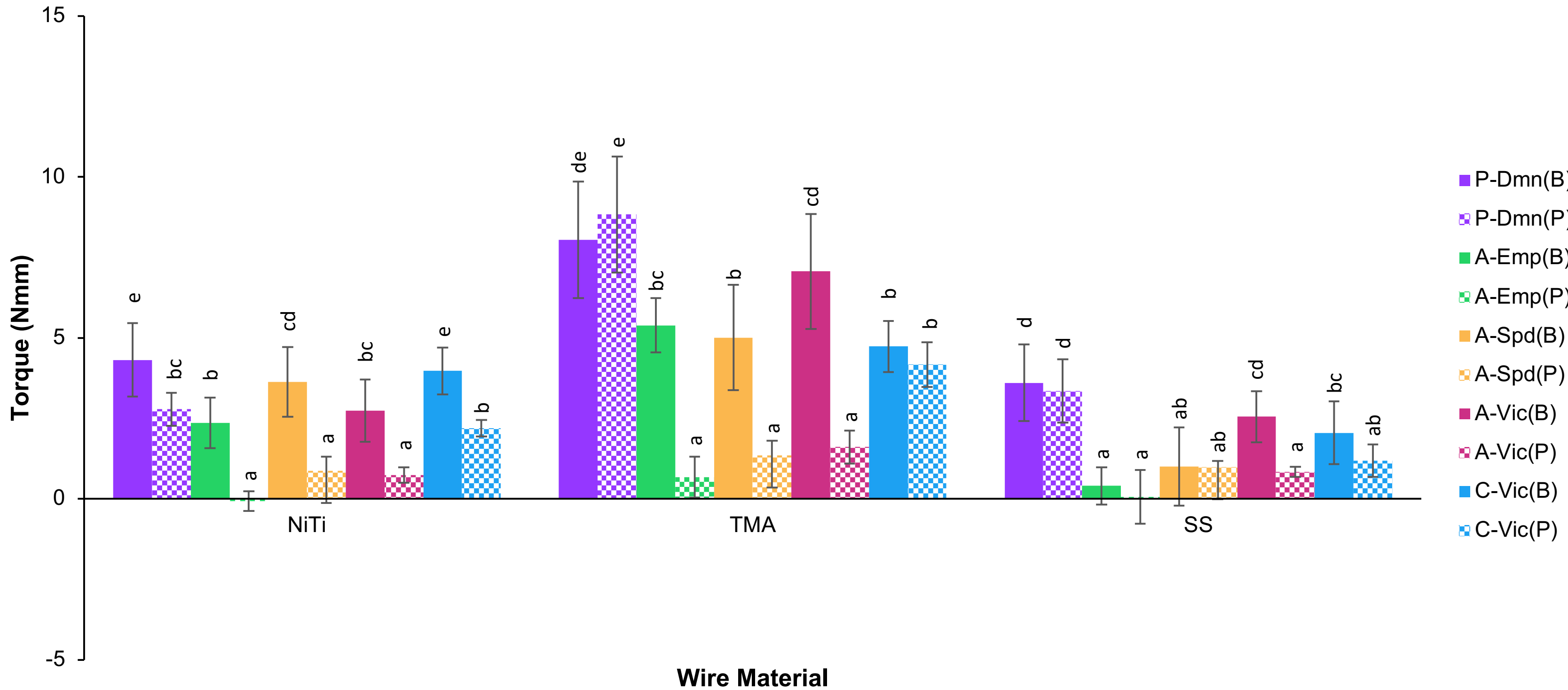

Figure 21: Mean torquing moments in Nmm (±SD) measured at 15' degrees with different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

4.1.2 Comparing Wire Materials

Comparing mean torquing moments generated by a given bracket system with different wire materials and degrees rotation showed a statistically significant difference for all bracket groups ($P<0.001$). Associated means (±SD) with significance results are found in both Table 6 and Figures 22-26, and associated P-values are included in Table 7.

Generally, NiTi wires produced significantly lower mean torquing moments than the other wire materials, regardless of bracket group and degree of rotation ($P<0.001$), with the exception of A-Vic(P) and C-Vic(B) groups at 15 degrees of rotation ($P=1.00$ and $P=0.189$ respectively).

With SS wires, no clear pattern emerged at 15 degrees of rotation. However, at 30 and 45 degrees of rotation mean torquing moments generated were significantly higher than the other wire materials, regardless of bracket group ($P<0.05$), apart from A-Vic(P) at 30 degrees ($P=0.474$).

For TMA wires, it was found to produce intermediate mean torquing moments, especially at higher degrees of rotation. Torquing moments were significantly higher than NiTi ($P<0.001$) and significantly lower than SS ($P<0.05$) at 30 and 45 degrees of rotation, with the exception of A-Vic(P) at 30 degrees of rotation with SS wires ($P=0.474$).

Table 6: Mean torquing moments in Nmm (±SD) generated for each bracket system-wire combination for every 15 degree increment of rotation for both loading and unloading, where unloading values are represented by prime (′) values. Non-significant differences between different wire materials within a bracket group-rotational increment at P>0.05 are denoted by shared alphabetical letters within each row

		Wire Material		
Bracket System	Degree	NiTi	TMA	SS
P-Dmn(B)	15	5.7(1.4)^{a}	11.1(2.6)^{b}	18.0(1.7)^{c}
	30	16.1(1.1)^{a}	34.1(4.2)^{b}	52.2(2.3)^{c}
	45	20.2(1.0)^{a}	56.9(3.4)^{b}	84.8(1.9)^{c}
	30'	10.7(0.9)^{a}	30.3(2.8)^{b}	35.5(1.6)^{c}
	15'	4.3(1.1)^{a}	8.0(1.8)^{b}	3.6(1.2)^{a}
P-Dmn(P)	15	3.4(0.7)^{a}	12.8(2.9)^{b}	14.1(2.7)^{b}
	30	13.2(1.4)^{a}	36.2(3.2)^{b}	47.7(4.7)^{c}
	45	19.3(0.8)^{a}	57.5(2.3)^{b}	79.9(3.8)^{c}
	30'	10.0(0.7)^{a}	31.3(2.2)^{b}	32.0(3.0)^{b}
	15'	2.8(0.5)^{a}	8.8(1.8)^{b}	3.4(1.0)^{a}
A-Emp(B)	15	3.8(0.8)^{a}	8.23(0.8)^{b}	8.2(2.3)^{b}
	30	15.3(0.8)^{a}	30.0(1.4)^{b}	39.3(3.0)^{c}
	45	20.8(0.4)^{a}	52.7(1.4)^{b}	72.5(2.3)^{c}
	30'	10.7(0.3)^{a}	26.5(1.0)^{c}	24.2(1.5)^{b}
	15'	2.4(0.8)^{b}	5.4(0.9)^{c}	0.4(0.6)^{a}
A-Emp(P)	15	0.9(0.4)^{a}	3.0(1.0)^{b}	2.8(0.6)^{b}
	30	8.8(1.2)^{a}	18.3(1.1)^{b}	23.9(1.0)^{c}
	45	17.0(1.4)^{a}	38.5(0.9)^{b}	46.6(1.2)^{c}
	30'	6.9(1.0)^{a}	14.9(0.8)^{c}	8.9(0.9)^{b}
	15'	-0.1(0.3)^{a}	0.7(0.6)^{b}	0.1(0.8)^{ab}
A-Spd(B)	15	4.6(1.3)^{a}	7.0(1.7)^{b}	8.1(2.5)^{b}
	30	15.5(1.3)^{a}	26.7(4.6)^{b}	38.8(2.9)^{c}
	45	20.8(0.7)^{a}	49.1(4.9)^{b}	64.6(14.2)^{c}
	30'	11.3(0.7)^{a}	24.5(4.2)^{b}	21.3(9.6)^{b}
	15'	3.6(1.1)^{b}	5.0(1.6)^{b}	1.0(1.2)^{a}
A-Spd(P)	15	1.2(0.2)^{a}	4.8(1.6)^{b}	3.7(1.2)^{b}
	30	7.3(0.9)^{a}	15.9(1.7)^{b}	18.8(3.1)^{c}
	45	15.1(0.8)^{a}	29.5(2.3)^{b}	34.3(4.4)^{c}
	30'	5.2(0.6)^{a}	10.2(1.4)^{c}	6.5(1.0)^{b}
	15'	0.8(0.4)^{a}	1.4(0.5)^{b}	1.0(0.2)^{ab}
A-Vic(B)	15	3.9(1.2)^{a}	10.4(2.3)^{b}	11.9(2.3)^{b}
	30	14.1(1.4)^{a}	31.5(3.4)^{b}	41.6(3.8)^{c}
	45	19.3(0.8)^{a}	54.0(2.3)^{b}	73.0(4.5)^{c}
	30'	10.2(0.6)^{a}	27.9(2.7)^{b}	26.5(2.8)^{b}
	15'	2.7(1.0)^{a}	7.1(1.8)^{b}	2.6(0.8)^{a}
A-Vic(P)	15	1.1(0.3)^{a}	3.0(1.4)^{b}	1.2(0.4)^{a}
	30	6.1(1.0)^{a}	17.9(2.5)^{b}	16.7(2.6)^{b}
	45	15.1(1.1)^{a}	37.6(3.3)^{b}	41.0(3.5)^{c}
	30'	5.0(0.8)^{a}	15.0(2.3)^{c}	8.1(1.3)^{b}
	15'	0.7(0.2)^{a}	1.6(0.5)^{b}	0.8(0.2)^{a}
C-Vic(B)	15	4.9(0.8)^{a}	6.9(1.4)^{a}	9.9(3.5)^{b}
	30	16.3(0.7)^{a}	29.5(1.6)^{b}	44.7(3.3)^{c}
	45	20.7(0.4)^{a}	53.7(1.2)^{b}	79.1(2.8)^{c}
	30'	11.1(0.3)^{a}	26.5(0.8)^{b}	28.2(2.0)^{c}
	15'	4.0(0.7)^{b}	4.7(0.8)^{b}	2.1(1.0)^{a}
C-Vic(P)	15	2.5(0.2)^{a}	6.3(0.9)^{b}	7.9(1.3)^{c}
	30	11.5(0.9)^{a}	27.6(1.8)^{b}	37.2(3.3)^{c}
	45	19.8(0.7)^{a}	51.5(2.3)^{b}	71.1(4.8)^{c}
	30'	9.5(0.7)^{a}	24.5(1.7)^{b}	23.6(2.8)^{b}
	15'	2.3(0.3)^{b}	4.2(0.7)^{c}	1.2(0.5)^{a}

Table 7: Significance results comparing mean torquing moments between wire materials within a bracket system-rotational increment

Wire Material	Degree	Significance Results
P-Dmn(B)	15	F(2, 27) = 104.87, P<0.001
	30	F(2, 27) = 419.26, P<0.001
	45	F(2, 27) = 1924.27, P<0.001
	30′	F(2, 27) = 470.27, P<0.001
	15′	F(2, 27) = 28.59, P<0.001
P-Dmn(P)	15	F(2, 27) = 63.64, P<0.001
	30	F(2, 27) = 270.78, P<0.001
	45	F(2, 27) = 1382.06, P<0.001
	30′	F(2, 27) = 334.15, P<0.001
	15′	F(2, 27) = 75.63, P<0.001
A-Emp(B)	15	F(2, 27) = 149.11, P<0.001
	30	F(2, 27) = 477.39, P<0.001
	45	F(2, 27) = 2270.44, P<0.001
	30′	F(2, 27) = 1505.68, P<0.001
	15′	F(2, 27) = 109.15, P<0.001
A-Emp(P)	15	F(2, 27) = 62.18, P<0.001
	30	F(2, 27) = 634.73, P<0.001
	45	F(2, 27) = 2946.63, P<0.001
	30′	F(2, 27) = 589.70, P<0.001
	15′	F(2, 27) = 5.05, P=0.014
A-Spd(B)	15	F(2, 27) = 9.64, P<0.001
	30	F(2, 27) = 133.45, P<0.001
	45	F(2, 27) = 59.17, P<0.001
	30′	F(2, 27) = 13.42, P<0.001
	15′	F(2, 27) = 24.02, P<0.001
A-Spd(P)	15	F(2, 27) = 24.98, P<0.001
	30	F(2, 27) = 77.51, P<0.001
	45	F(2, 27) = 119.12, P<0.001
	30′	F(2, 27) = 63.56, P<0.001
	15′	F(2, 27) = 4.45, P=0.021
A-Vic(B)	15	F(2, 27) = 44.51, P<0.001
	30	F(2, 27) = 209.10, P<0.001
	45	F(2, 27) = 860.55, P<0.001
	30′	F(2, 27) = 194.10, P<0.001
	15′	F(2, 27) = 41.53, P<0.001
A-Vic(P)	15	F(2, 27) = 15.39, P<0.001
	30	F(2, 27) = 90.06, P<0.001
	45	F(2, 27) = 252.12, P<0.001
	30′	F(2, 27) = 100.90, P<0.001
	15′	F(2, 27) = 20.38, P<0.001
C-Vic(B)	15	F(2, 27) = 13.25, P<0.001
	30	F(2, 27) = 358.50, P<0.001
	45	F(2, 27) = 2697.46, P<0.001
	30′	F(2, 27) = 585.52, P<0.001
	15′	F(2, 27) = 28.15, P<0.001
C-Vic(P)	15	F(2, 27) = 85.61, P<0.001
	30	F(2, 27) = 343.39, P<0.001
	45	F(2, 27) = 700.88, P<0.001
	30′	F(2, 27) = 191.12, P<0.001
	15′	F(2, 27) = 88.47, P<0.001

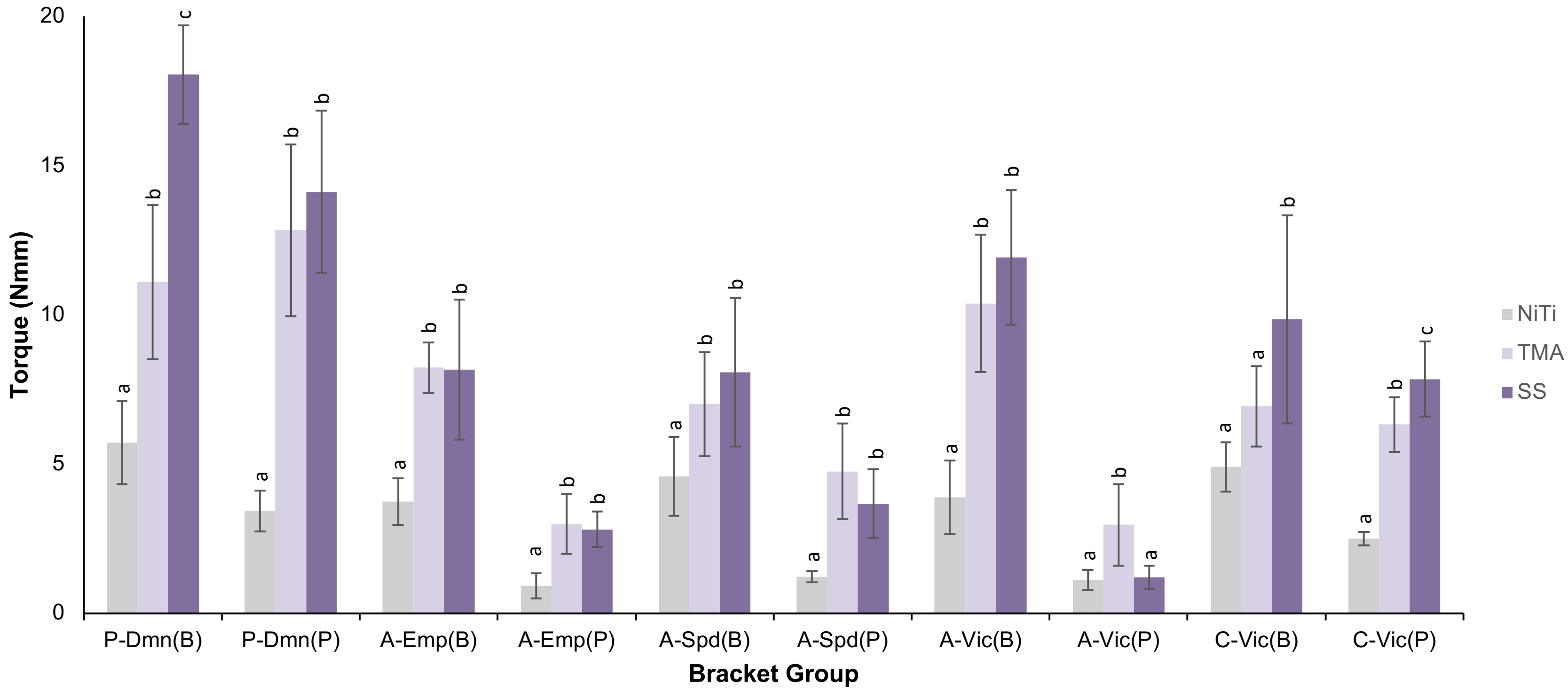

Figure 22: Mean torquing moments in Nmm (±SD) measured at 15 degrees with different wire materials versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

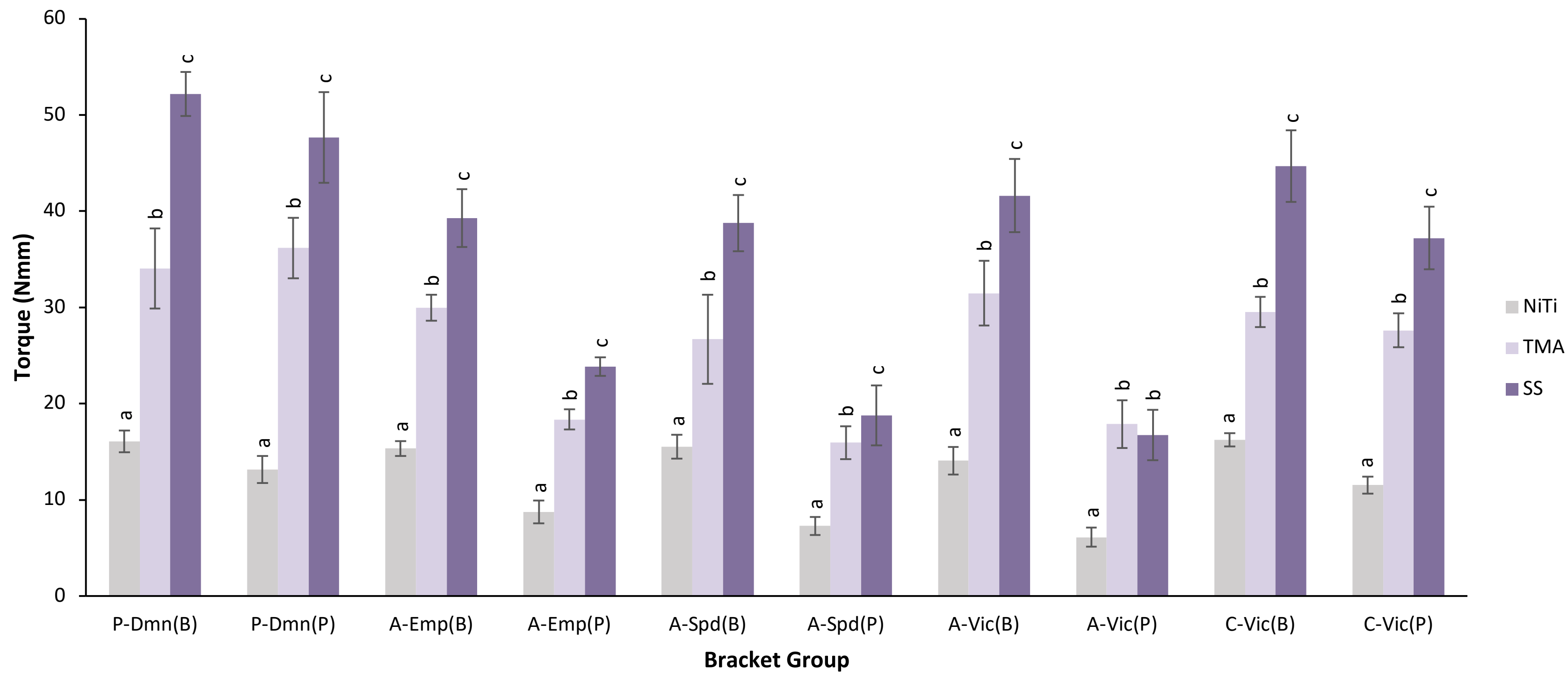

Figure 23: Mean torquing moments in Nmm (±SD) measured at 30 degrees with different wire materials versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

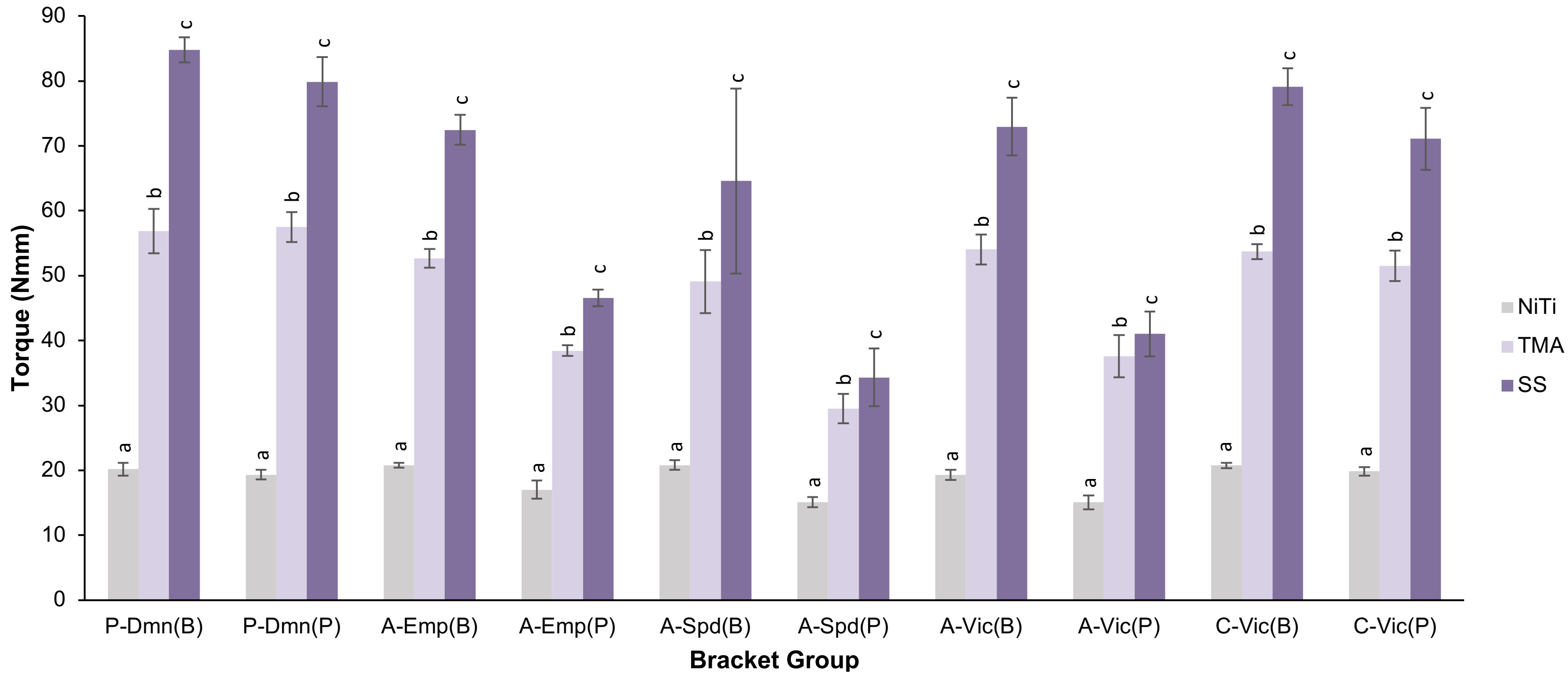

Figure 24: Mean torquing moments in Nmm (±SD) measured at 45 degrees with different wire materials versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

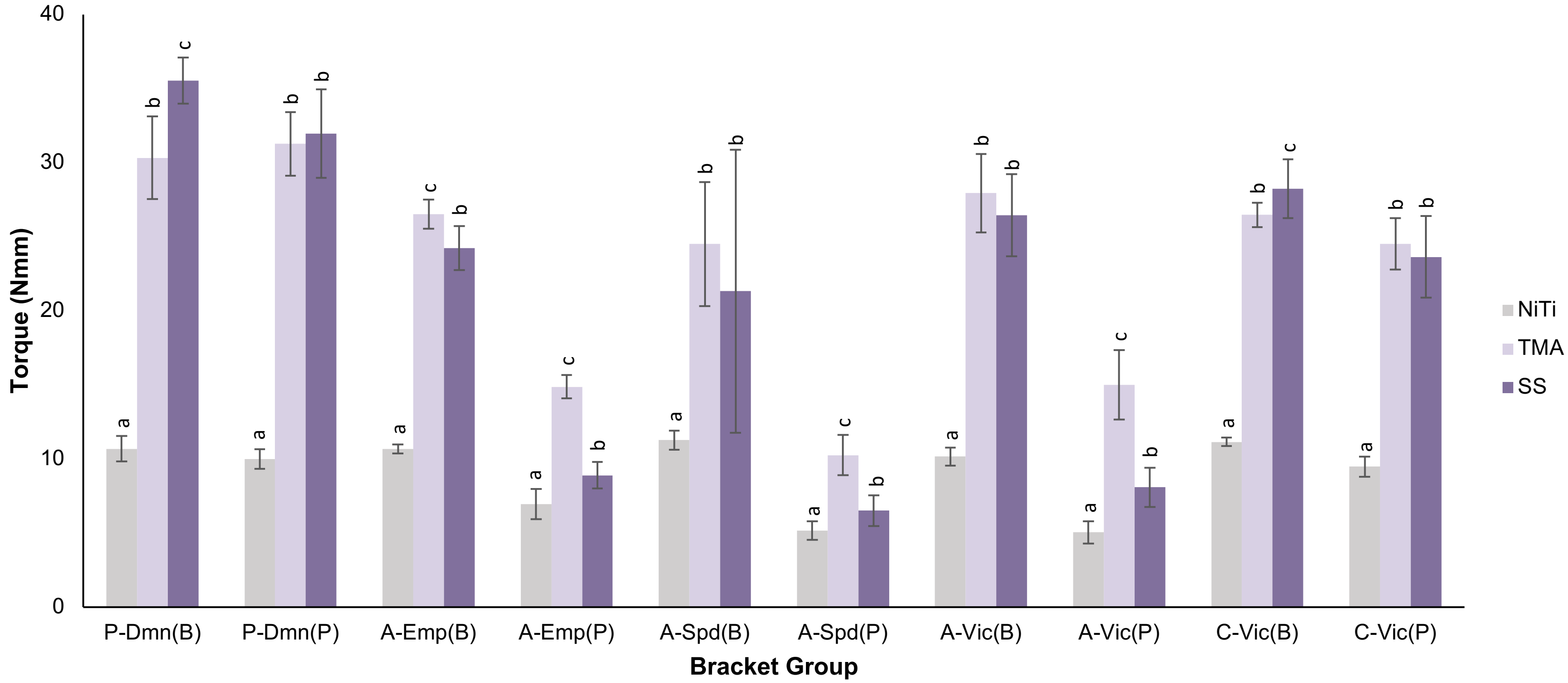

Figure 25: Mean torquing moments in Nmm (±SD) measured at 30' degrees with different wire materials versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

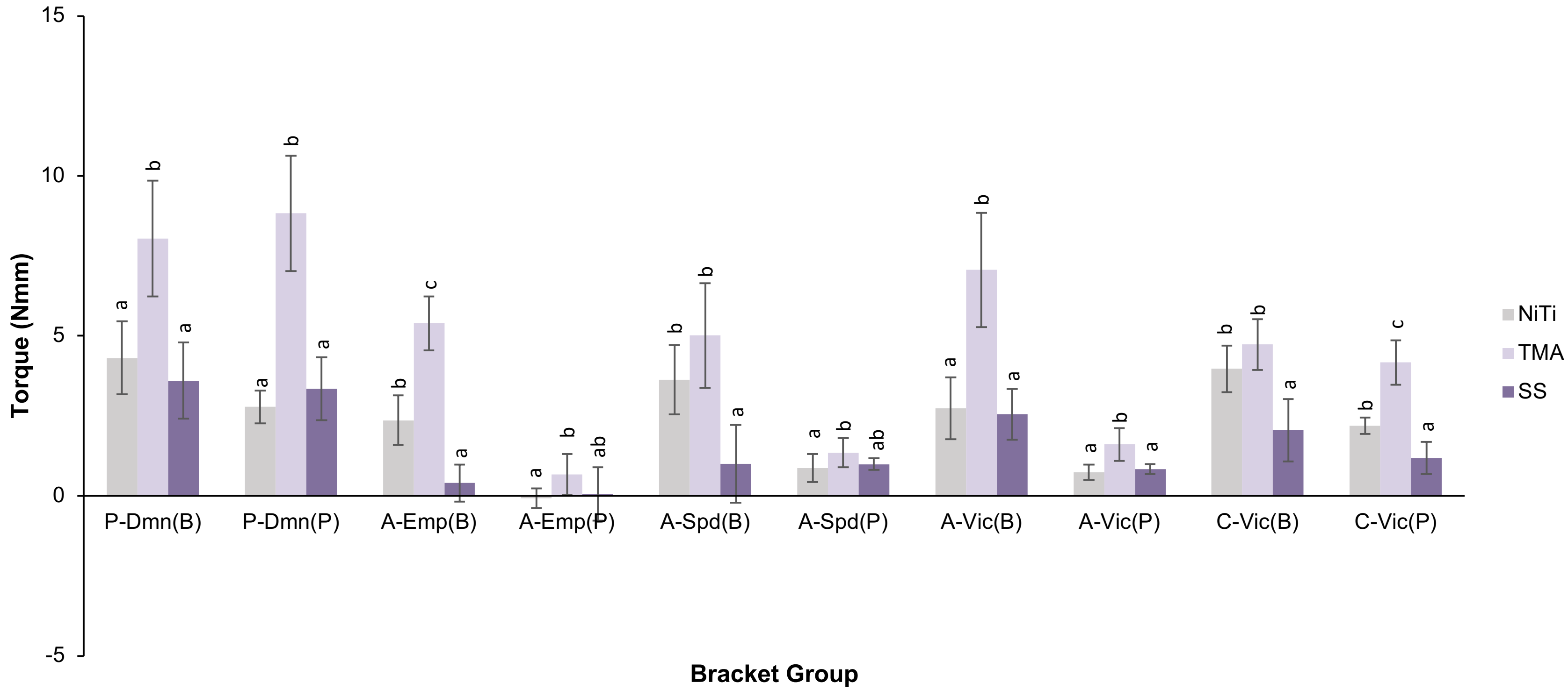

Figure 26: Mean torquing moments in Nmm (±SD) measured at 15' degrees with different wire materials versus bracket system. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

4.2 Torquing Direction

Comparing each bracket system in both buccal and palatal root torque simulations with different wire materials showed that direction of torque simulation tended to influence ASL systems more than PSL and conventionally ligated bracket systems. This is shown in Figures 27-31 and Table 8. Associated significance results are shown in Table 4, Table 5, and Figures 17-21.

Examining the results further, in the ASL bracket groups there are significant differences in torquing moments generated between buccal and palatal root torque simulations, at each 15 degree loading interval, for most wire groups, with the exception of the TMA wire at 15 degrees of rotation with the A-Spd brackets ($P=0.176$). In all other cases the ASL bracket systems generated significantly larger torquing moments in buccal root torque compared to palatal root torque simulations ($P<0.001$). For example, considering SS wires at 45 degrees of rotation, the A-Emp (72.5 vs 46.6 Nmm), A-Spd (64.6 vs 34.3 Nmm), and A-Vic (73.0 vs 41.0 Nmm) groups produced significantly greater torquing moments in the buccal root torque direction compared to palatal root torque ($P<0.001$), while the P-Dmn (84.8 vs 79.9 Nmm) and C-Vic (79.1 vs 71.1 Nmm) groups did not ($P=1.00$ and $P=0.111$, respectively).

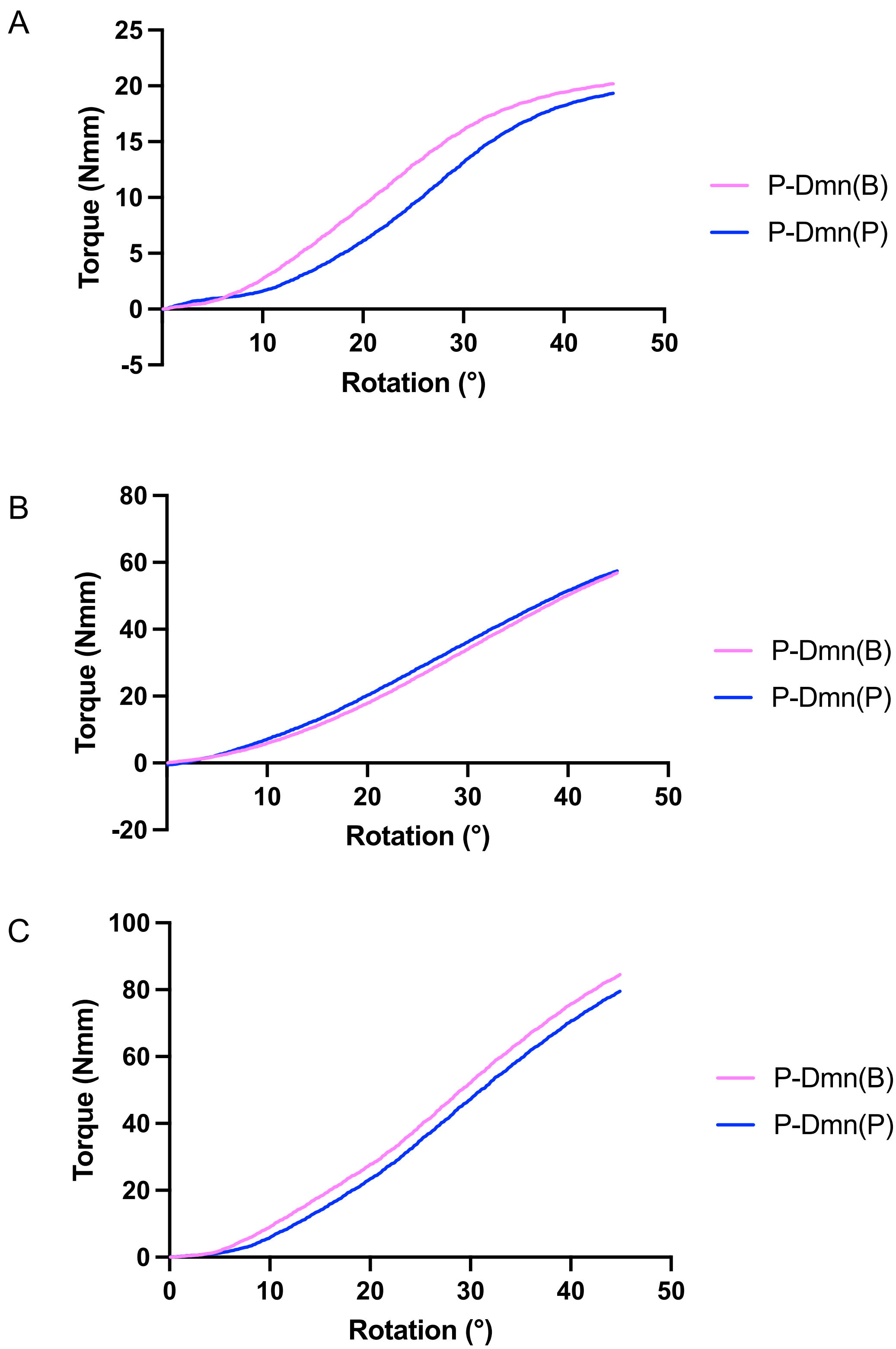

Figure 27: Mean torquing moments measured for P-Dmn brackets in both buccal (P-Dmn(B)) and palatal (P-Dmn(P)) root torque simulation directions with NiTi (A), TMA (B), and SS (C) wires

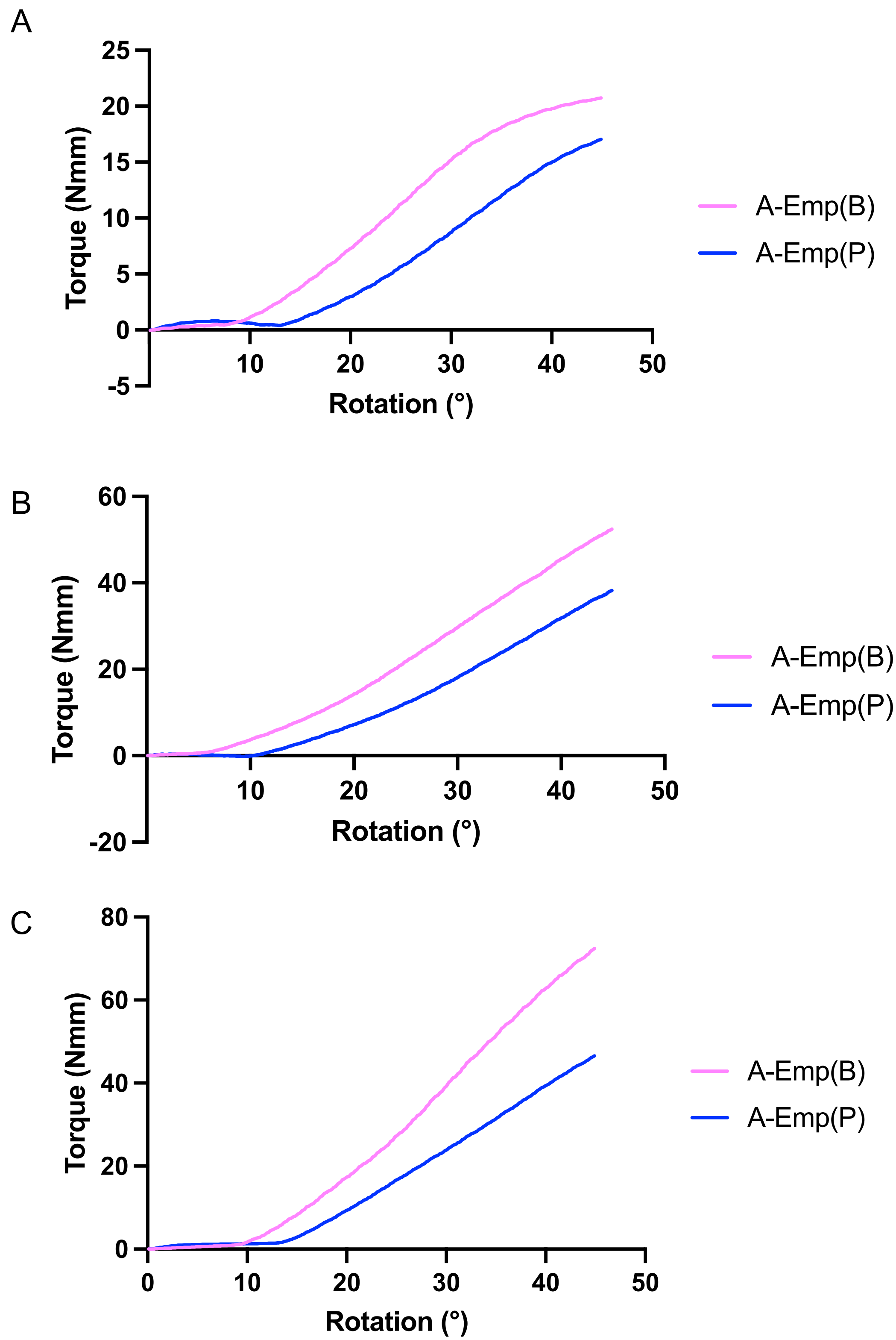

Figure 28: Mean torquing moments measured for A-Emp brackets in both buccal (A-Emp(B)) and palatal (A-Emp(P)) root torque simulation directions with NiTi (A), TMA (B), and SS (C) wires

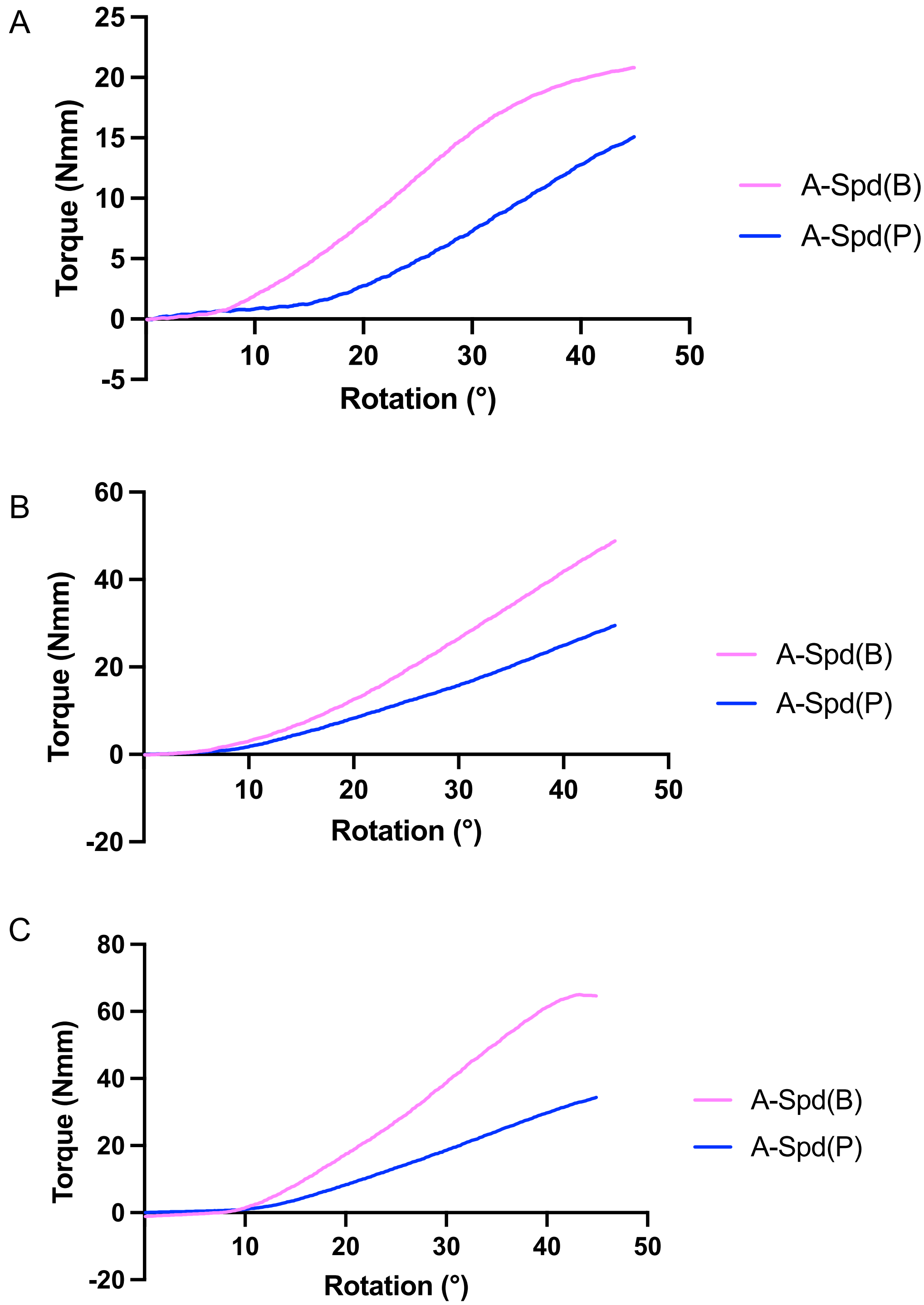

Figure 29: Mean torquing moments measured for A-Spd brackets in both buccal (A-Spd(B)) and palatal (A-Spd(P)) root torque simulation directions with NiTi (A), TMA (B), and SS (C) wires

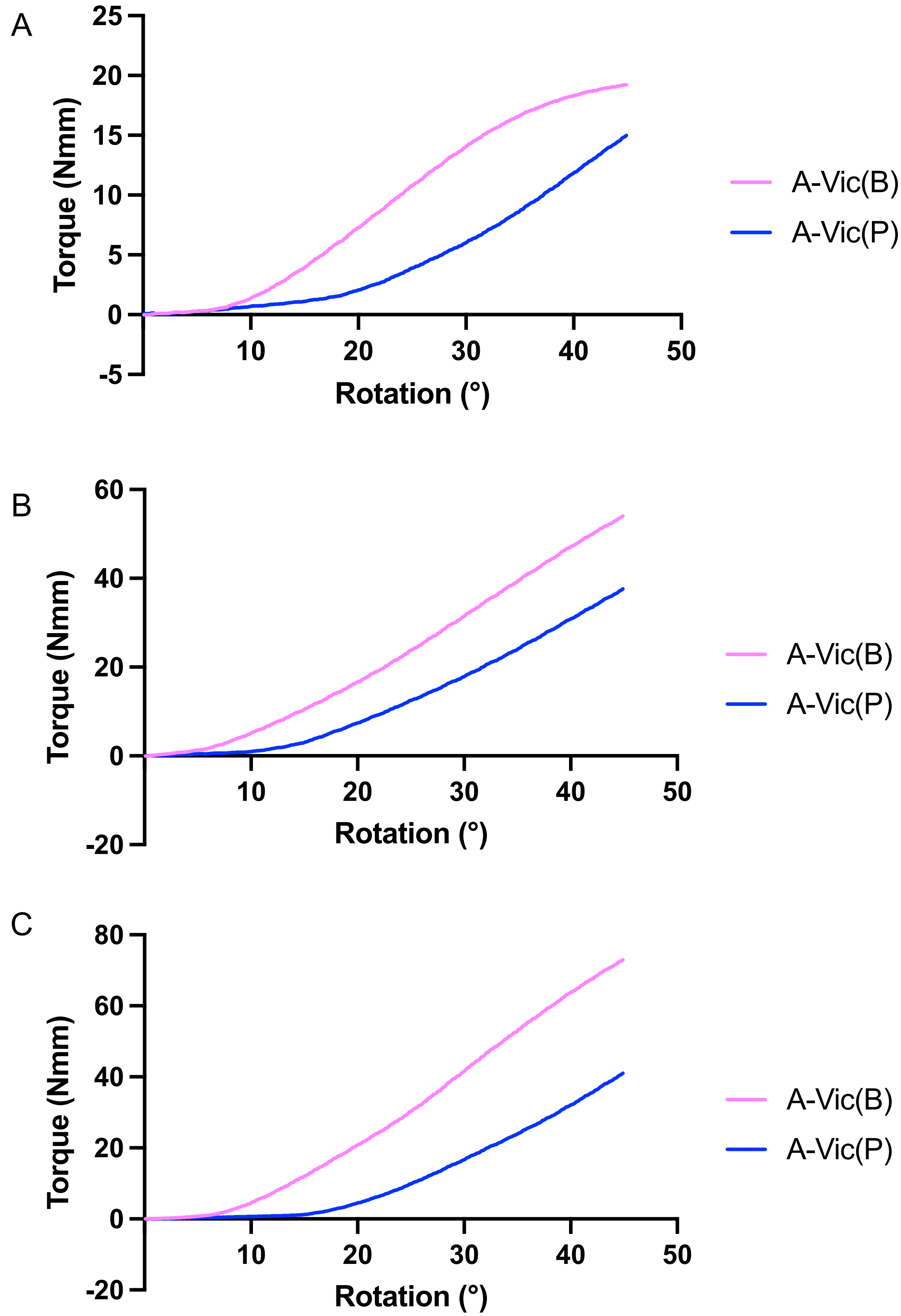

Figure 30: Mean torquing moments measured for A-Vic brackets in both buccal (A-Vic(B)) and palatal (A-Vic(P)) root torque simulation directions with NiTi (A), TMA (B), and SS (C) wires

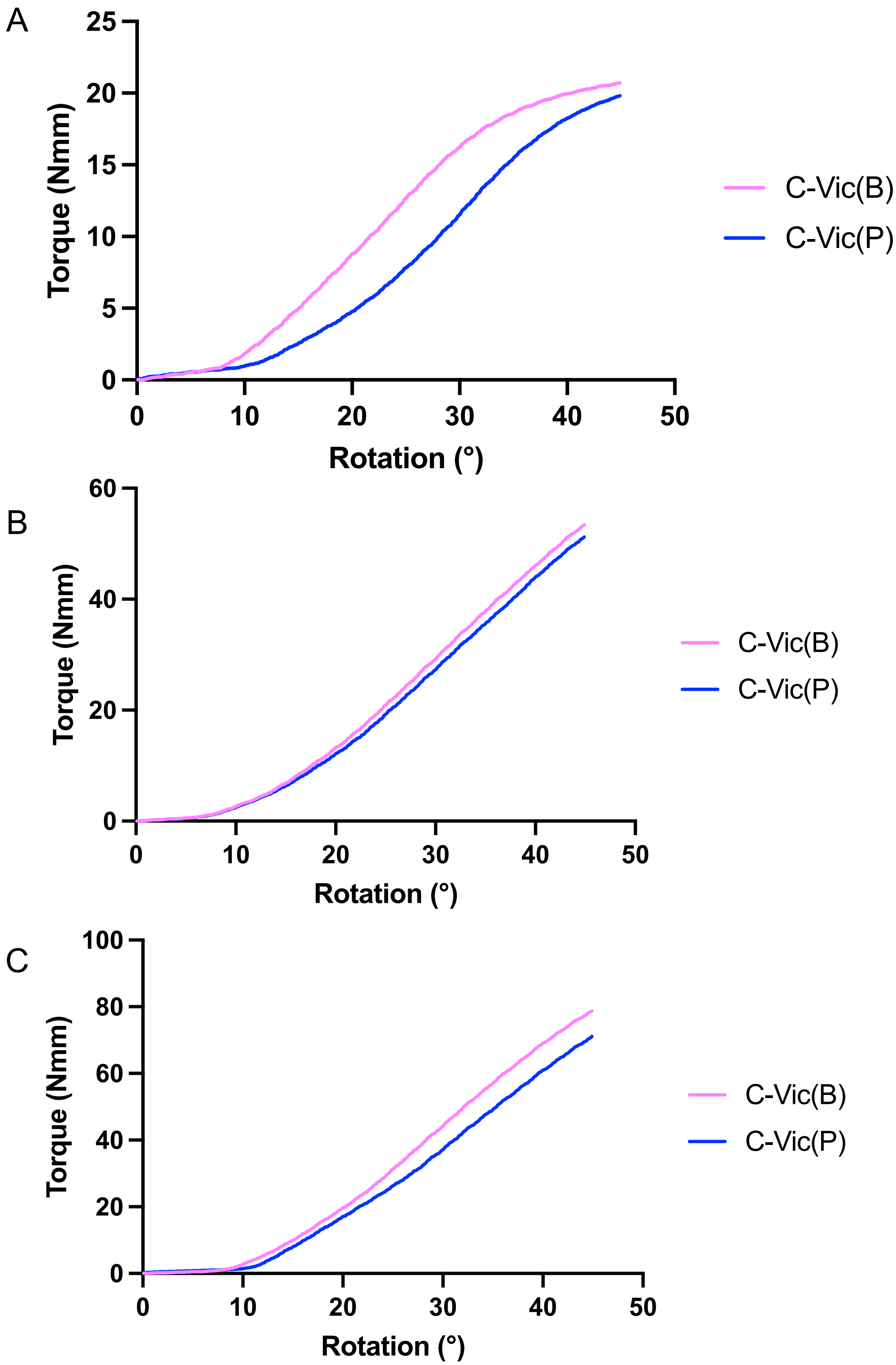

Figure 31: Mean torquing moments measured for C-Vic brackets in both buccal (C-Vic(B)) and palatal (C-Vic(P)) root torque simulation directions with NiTi (A), TMA (B), and SS (C) wires

Table 8: Comparison of mean torquing moments (±SD) generated in buccal and palatal root torque directions for each bracket system-wire combination, at every 15 degree increment of the loading curves. Significant differences between rotation direction within a bracket system-wire material rotational increment at P<0.05 are denoted by (*)

		Wire Material		
Bracket System	**Degrees**	NiTi	TMA	SS
		Buccal / Palatal	Buccal / Palatal	Buccal / Palatal
P-Dmn	15	5.7(1.4) / 3.4(0.7)*	11.1(2.6) / 12.8(2.9)	18.0(1.7) / 14.1(2.7)*
	30	16.1(1.1) / 13.2(1.4)*	34.1(4.2) / 36.2(3.2)	52.2(2.3) / 47.7(4.7)*
	45	20.2(1.0) / 19.3(0.8)	56.9(3.4) / 57.5(2.3)	84.8(1.9) / 79.9(3.8)
A-Emp	15	3.8(0.8) / 0.9(0.4)*	8.2(0.8) / 3.0(1.0)*	8.2(2.3) / 2.8(0.6)*
	30	15.3(0.8) / 8.8(1.2)*	30.0(1.4) / 18.3(1.1)*	39.3(3.0) / 23.9(1.0)*
	45	20.8(0.4) / 17.0(1.4)*	52.7(1.4) / 38.5(0.9)*	72.5(2.3) / 46.6(1.3)*
A-Spd	15	4.6(1.3) / 1.2(0.2)*	7.0(1.7) / 4.8(1.6)	8.1(2.5) / 3.7(1.2)*
	30	15.5(1.3) / 7.3(0.9)*	26.7(4.6) / 15.9(1.7)*	38.8(2.9) / 18.8(3.1)*
	45	20.8(0.7) / 15.1(0.8)*	49.1(4.9) / 29.5(2.3)*	64.6(14.2) / 34.3(4.4)*
A-Vic	15	3.9(1.2) / 1.1(0.3)*	10.4(2.3) / 3.0(1.4)*	11.9(2.3) / 1.2(0.4)*
	30	14.1(1.4) / 6.1(1.0)*	31.5(3.4) / 17.9(2.5)*	41.6(3.8) / 16.7(2.6)*
	45	19.3(0.8) / 15.1(1.1)*	54.0(2.3) / 37.6(3.3)*	73.0(4.5) / 41.0(3.5)*
C-Vic(B)	15	4.9(0.8) / 2.5(0.2)*	6.9(1.4) / 6.3(0.9)	9.9(3.5) / 7.9(1.3)
	30	16.3(0.7) / 11.5(0.9)*	29.5(1.6) / 27.6(1.8)	44.7(3.7) / 37.2(3.3)*
	45	20.7(0.4) / 19.8(0.7)	53.7(1.2) / 51.5(2.3)	79.1(2.8) / 71.1(4.8)

4.3 Engagement Angles

Comparing mean engagement angles recorded for different bracket groups within a given wire material revealed significant differences between groups for all comparisons ($P<0.001$). Associated mean engagement angles (±SD) along with significance results are highlighted in Table 9 and Figure 32, with associated P-values found in Table 10.

Across all wire materials, bracket groups in the palatal root torque simulation generated significantly larger engagement angles than their buccal root torque counterparts. Two exceptions were noted, with P-Dmn and A-Spd groups in TMA wires, although these differences were not significant ($P=1.00$). It was also noted that the P-Dmn(P) group generated significantly smaller engagement angles for all wire materials in both directions of rotation ($P<0.001$), with the exception of C-Vic(P) in NiTi wires ($P=0.186$).

Comparing mean engagement angles between wire materials within each bracket system revealed a significant difference for all groups ($P<0.05$), except A-Spd(B) and C-Vic(B) ($P=0.314$ and $P=0.633$, respectively). Associated mean engagement angles (±SD) along with significance results are highlighted in Table 11 and Figure 33, with associated P-values found in Table 12.

When considering comparisons between wire materials within each bracket group, TMA produced significantly smaller engagement angles ($P<0.05$) in the majority of bracket systems. NiTi produced significantly larger engagement angles ($P<0.05$) in the majority of bracket groups, and SS was found to produced intermediate angles of engagement.

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

	Bracket Group									
Wire Material	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)
NiTi	6.9(2.5)[a]	11.6(1.7)[bc]	10.7(1.6)[b]	16.2(1.4)[d]	9.3(2.5)[ab]	16.3(2.5)[d]	9.5(2.2)[ab]	20.1(2.3)[e]	8.7(1.4)[ab]	14.3(1.6)[cd]
TMA	5.2(2.0)[ab]	4.1(2.4)[a]	7.3(1.0)[bc]	12.7(1.6)[d]	8.9(2.6)[c]	8.7(3.2)[c]	5.6(2.2)[ab]	12.7(2.3)[d]	9.2(1.7)[c]	9.5(1.1)[cd]
SS	4.7(0.9)[a]	6.4(1.3)[ab]	10.2(1.4)[de]	13.2(0.5)[fg]	10.4(1.7)[de]	11.7(1.5)[ef]	7.4(1.2)[bc]	17.1(1.5)[g]	9.4(2.1)[cd]	10.4(0.7)[de]

Table 10: Significance results comparing mean loading curve engagement angles between bracket systems within a wire material

Wire Material	Significance Results
NiTi	$F(9,90) = 43.33, P<0.001$
TMA	$F(9,90) = 19.28, P<0.001$
SS	$F(9,90) = 67.62, P<0.001$

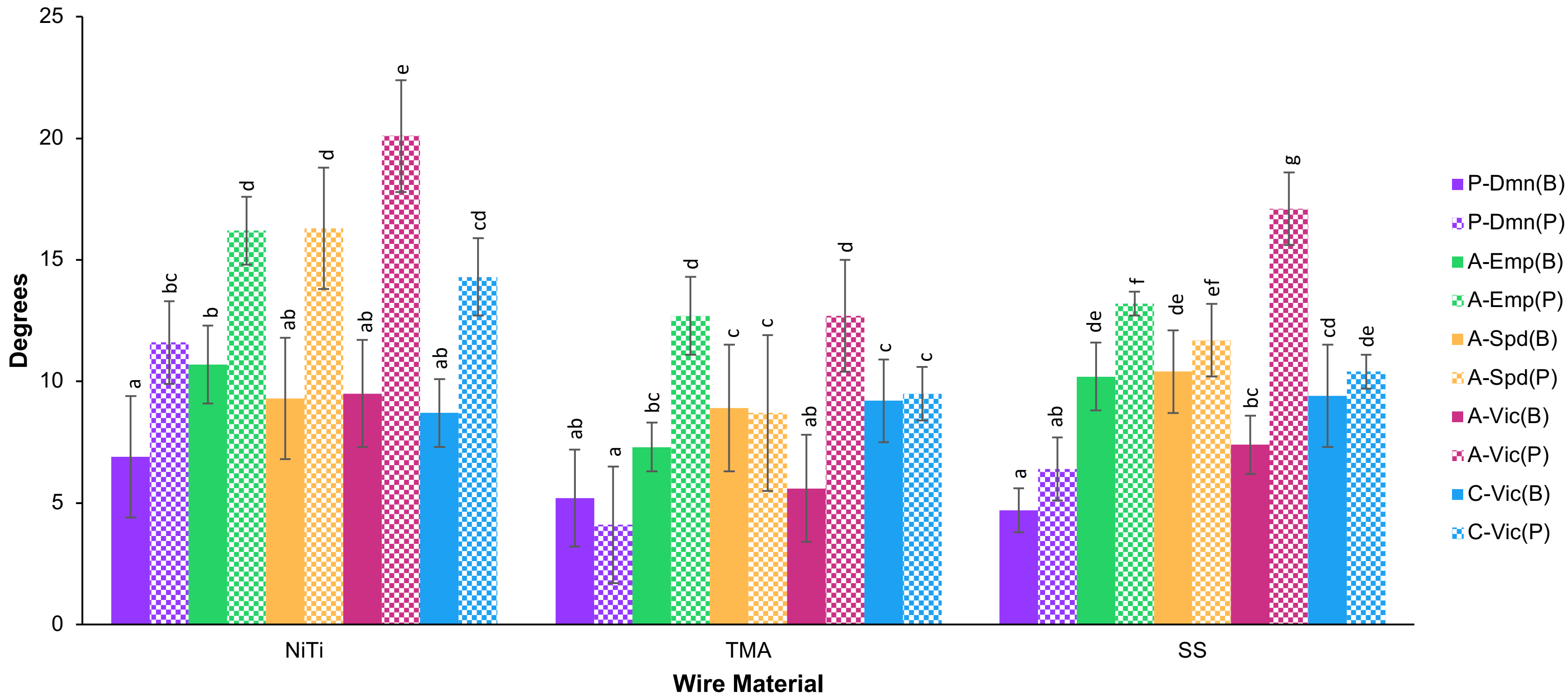

Figure 32: Mean loading curve engagement angles in degrees (±SD) with different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

Table 11: Mean loading curve engagement angles in degrees (±SD) for each bracket system-wire combination. Non-significant differences between wire materials within a given bracket group at P>0.05 are denoted by shared alphabetical letters within each row

	Wire Material		
Bracket Group	NiTi	TMA	SS
P-Dmn(B)	6.9(2.5)[b]	5.2(2.0)[ab]	4.7(0.9)[a]
P-Dmn(P)	11.6(1.7)[c]	4.1(2.4)[a]	6.4(1.3)[b]
A-Emp(B)	10.7(1.6)[b]	7.3(1.0)[a]	10.2(1.4)[b]
A-Emp(P)	16.2(1.4)[b]	12.7(1.6[a])	13.2(0.5)[a]
A-Spd(B)	9.3(2.5)[a]	8.9(2.6)[a]	10.4(1.7)[a]
A-Spd(P)	16.3(2.5)[c]	8.7(3.2)[a]	11.7(1.5)[b]
A-Vic(B)	9.5(2.2)[b]	5.6(2.2)[a]	7.4(1.2)[ab]
A-Vic(P)	20.1(2.3)[c]	12.7(2.3)[a]	17.1(1.5)[b]
C-Vic(B)	8.7(1.4)[a]	9.2(1.7)[a]	9.4(2.1)[a]
C-Vic(P)	14.3(1.6)[b]	9.5(1.1)[a]	10.4(0.7)[a]

Table 12: Significance results comparing mean loading curve engagement angles between wires within a given bracket systems

Bracket Group	Significance Results
P-Dmn(B)	$F(2,27) = 3.50$, $P=0.045$
P-Dmn(P)	$F(2,27) = 42.61$, $P<0.001$
A-Emp(B)	$F(2,27) = 17.85$, $P<0.001$
A-Emp(P)	$F(2,27) = 22.61$, $P<0.001$
A-Spd(B)	$F(2,27) = 1.21$, $P=0.314$
A-Spd(P)	$F(2,27) = 23.09$, $P<0.001$
A-Vic(B)	$F(2,27) = 10.52$, $P<0.001$
A-Vic(P)	$F(2,27) = 31.22$, $P<0.001$
C-Vic(B)	$F(2,27) = 0.47$, $P=0.633$
C-Vic(P)	$F(2,27) = 46.59$, $P<0.001$

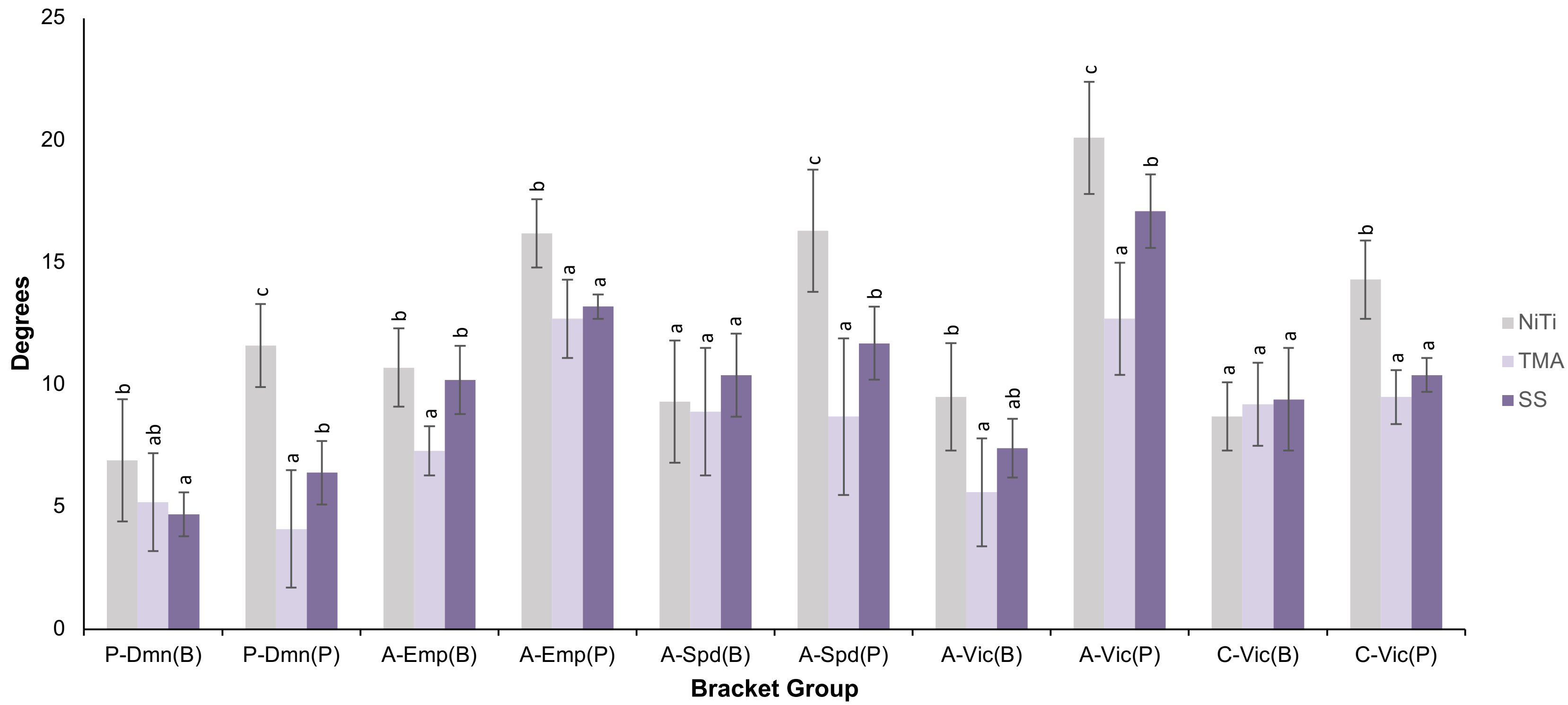

Figure 33: Mean loading curve engagement angles in degrees (±SD) with different wire materials versus bracket systems. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wires at P>0.05

4.4 Torsional Stiffness

Comparing the torsional stiffness recorded for different bracket groups within a given wire material showed that there were significant differences between the groups ($P<0.001$). This is shown in Figure 34 and Table 13. Associated significance results are shown in Table 14.

In SS wires, it was noted that in the palatal root torque direction A-Spd and A-Vic produced significantly smaller torsional stiffness values than the other bracket groups ($P<0.001$ and $P<0.05$, respectively). In the buccal root torque direction, A-Spd was found to produce the lowest torsional stiffness ($P<0.001$). Also in SS wires and ASL bracket groups, torsional stiffness values were significantly smaller in the palatal root torque direction when compared with the buccal root torque direction ($P<0.001$). With NiTi and TMA wires generally there were no significant differences in torsional stiffness values.

In comparing the torsional stiffness between wire materials within each bracket group, it was also shown that there were significant differences for all comparisons ($P<0.001$). This is demonstrated in Figure 35 and Table 15. Associated significance results are shown in Table 16.

The smallest torsional stiffness values were produced with NiTi wires, followed intermediately by TMA, with SS producing the highest torsional stiffness values. This pattern for significant ($P<0.001$) for all bracket groups, regardless of direction or rotation or ligation method.

(Note: Torsional stiffness values recorded for the loading curve, as measured between 20 and 25 degrees of rotation, and associated significance values, can be found in Appendices M-P.)

Table 13: Mean torsional stiffness in Nmm/° (±SD) for each bracket system-wire combination. Non-significant differences between bracket groups within a wire material at P>0.05 are denoted by shared alphabetical letters within each row

	Bracket Group									
Wire Material	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)
NiTi	0.9(0.02)^{abc}	0.8(0.02)^{ab}	0.9(0.03)^{abc}	0.8(0.06)^{a}	0.9(0.02)^{abc}	0.8(0.06)^{ab}	0.8(0.03)^{ab}	0.9(0.04)^{abc}	0.9(0.03)^{bc}	0.9(0.04)^{c}
TMA	1.8(0.05)^{ef}	1.8(0.06)^{ef}	1.8(0.03)^{cde}	1.7(0.03)^{b}	1.7(0.04)^{bc}	1.6(0.09)^{a}	1.8(0.06)^{de}	1.7(0.05)^{bcd}	1.9(0.06)^{f}	1.8(0.05)^{ef}
SS	3.5(0.04)^{de}	3.4(0.1)^{cd}	3.5(0.05)^{de}	3.2(0.09)^{c}	3.2(0.33)^{c}	2.6(0.3)^{a}	3.5(0.11)^{d}	2.9(0.11)^{b}	3.7(0.07)^{e}	3.6(0.08)^{de}

Table 14: Significance results comparing mean torsional stiffness between bracket systems within a wire material

Wire Material	Significance Results
NiTi	$F(9,90) = 4.30$, $P<0.001$
TMA	$F(9,90) = 26.66$, $P<0.001$
SS	$F(9,90) = 45.55$, $P<0.001$

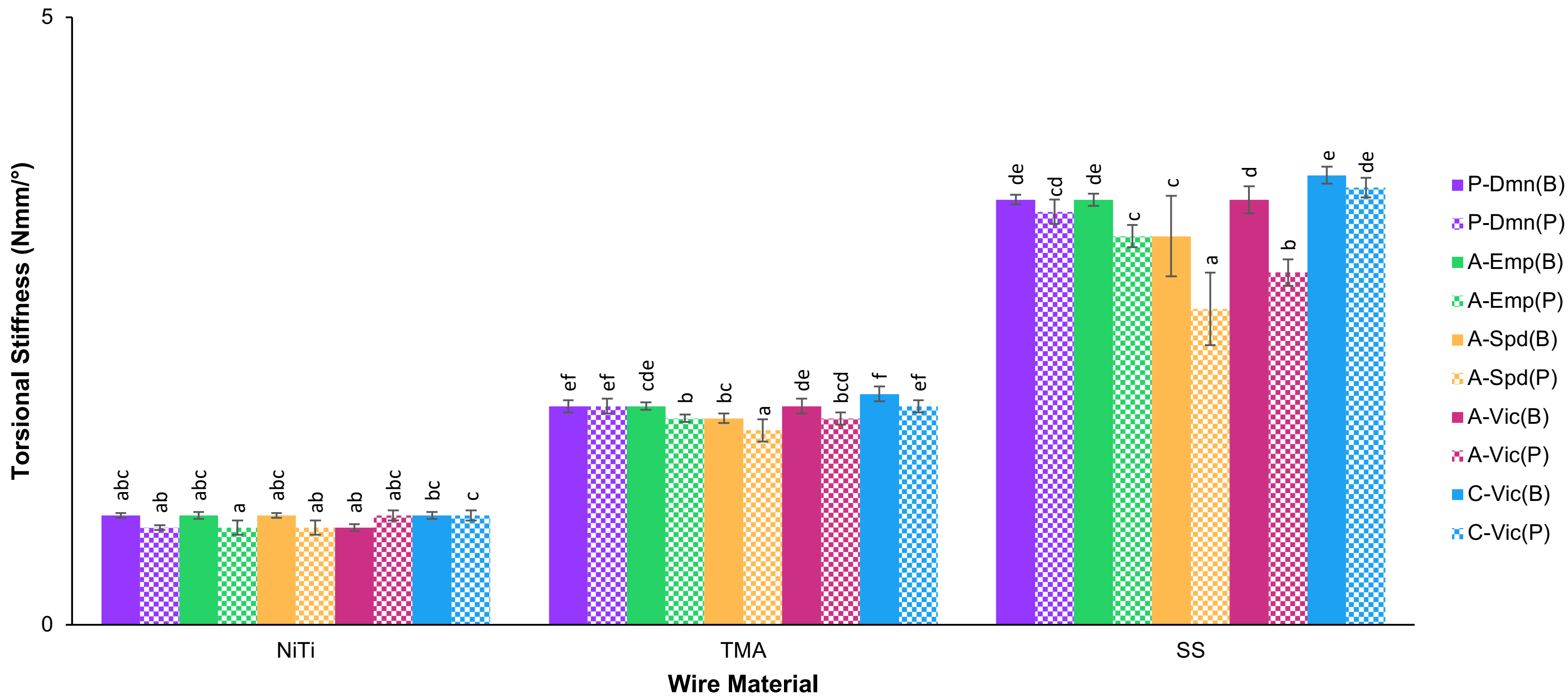

Figure 34: Mean torsional stiffness in Nmm/° (±SD) with different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

Table 15: Mean torsional stiffness in Nmm/° (±SD) for each bracket system-wire combination. Non-significant differences between wire materials within a given bracket group at P>0.05 are denoted by shared alphabetical letters within each row

	Wire Material		
Bracket Group	NiTi	TMA	SS
P-Dmn(B)	0.9(0.02)[a]	1.8(0.05)[b]	3.5(0.04)[c]
P-Dmn(P)	0.8(0.02)[a]	1.8(0.06)[b]	3.4(0.1)[c]
A-Emp(B)	0.9(0.03)[a]	1.8(0.03)[b]	3.5(0.05)[c]
A-Emp(P)	0.8(0.06)[a]	1.7(0.03)[b]	3.2(0.09)[c]
A-Spd(B)	0.9(0.02)[a]	1.7(0.04)[b]	3.2(0.33)[c]
A-Spd(P)	0.8(0.06)[a]	1.6(0.09)[b]	2.6(0.3)[c]
A-Vic(B)	0.8(0.03)[a]	1.8(0.06)[b]	3.5(0.11)[c]
A-Vic(P)	0.9(0.04)[a]	1.7(0.05)[b]	2.9(0.11)[c]
C-Vic(B)	0.9(0.03)[a]	1.9(0.06)[b]	3.7(0.07)[c]
C-Vic(P)	0.9(0.04)[a]	1.8(0.05)[b]	3.6(0.08)[c]

Table 16: Significance results comparing mean torsional stiffness between wire materials within a given bracket group

Bracket Group	Significance Results
P-Dmn(B)	$F(2,27) = 12549.87$, $P<0.001$
P-Dmn(P)	$F(2,27) = 3601.96$, $P<0.001$
A-Emp(B)	$F(2,27) = 11427.60$, $P<0.001$
A-Emp(P)	$F(2,27) = 3503.19$, $P<0.001$
A-Spd(B)	$F(2,27) = 366.16$, $P<0.001$
A-Spd(P)	$F(2,27) = 233.73$, $P<0.001$
A-Vic(B)	$F(2,27) = 3400.05$, $P<0.001$
A-Vic(P)	$F(2,27) = 2028.05$, $P<0.001$
C-Vic(B)	$F(2,27) = 6481.21$, $P<0.001$
C-Vic(P)	$F(2,27) = 5462.42$, $P<0.001$

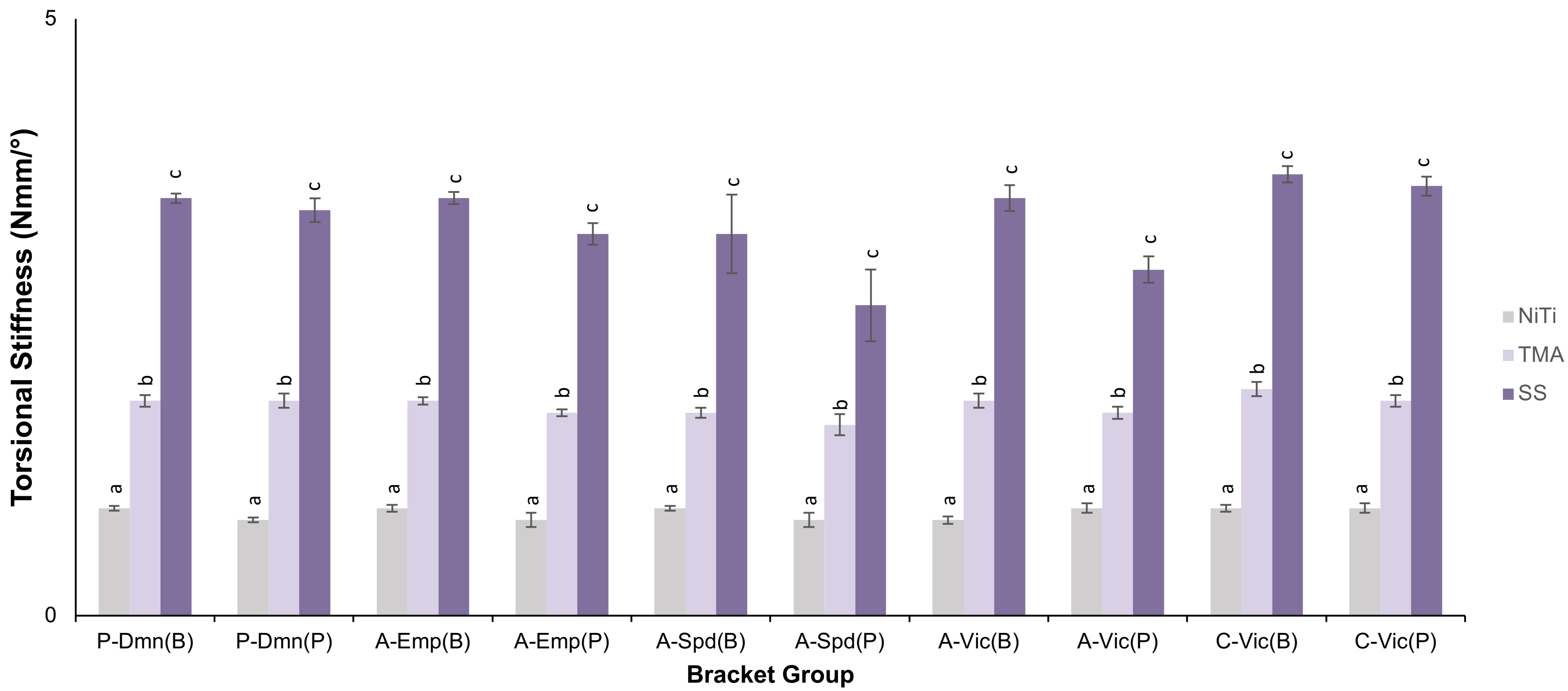

Figure 35: Mean torsional stiffness in Nmm/° (±SD) with different wire materials versus bracket systems. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wire materials at P>0.05

4.5 Hysteresis

Comparing the mean hysteresis recorded for different bracket groups within a given wire material showed that there were significant differences between groups ($P<0.001$). This is shown in Table 17 and Figure 36, with associated significance results shown in Table 18. Mean torque-rotation loading and unloading curves for all bracket systems and wire material combinations can be found in Appendices Q-U.

Across all bracket groups in NiTi wires, a pattern was noted in which hysteresis values were significantly higher when systems were rotated in a palatal root torque direction compared to a buccal root torque direction ($P<0.001$). In TMA and SS wires, hysteresis values were not significantly different across the majority of bracket groups.

There were also significant differences of mean hysteresis values between wire materials within each bracket group ($P<0.001$). This can be found in Table 19 and Figure 37, with associated significance results shown in Table 20.

In regards to wire materials, SS produced significantly higher hysteresis values, regardless of direction of rotation or bracket group ($P<0.001$). When examining ASL brackets, it was found that in the palatal root torque direction NiTi wires produced significantly lower hysteresis values ($P<0.01$).

Table 17: Mean hysteresis values in Nmm (±SD) for each bracket system-wire combination. Non-significant differences between bracket groups within a wire material at P>0.05 are denoted by shared alphabetical letters within each row

	Bracket System									
Wire Material	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)
NiTi	5.3(0.5)^{f}	3.1(1.1)^{bc}	4.4(0.6)^{def}	1.7(0.4)^{ab}	4.2(0.9)^{de}	2.0(0.4)^{b}	3.8(1.0)^{cd}	0.9(0.3)^{a}	5.0(0.5)^{ef}	1.9(0.3)^{b}
TMA	3.5(1.5)^{cd}	4.7(1.2)^{de}	2.8(0.5)^{bc}	2.9(0.5)^{bc}	1.4(0.7)^{a}	5.4(0.5)^{e}	3.3(0.9)^{bc}	2.7(0.7)^{abc}	2.2(1.1)^{ab}	2.3(0.4)^{abc}
SS	15.7(1.3)^{b}	14.4(2.2)^{b}	14.7(2.1)b	14.7(0.9)^{b}	16.8(9.5)^{b}	12.0(2.2)^{ab}	14.8(1.6)^{b}	8.5(1.7)^{a}	15.3(2.5)^{b}	13.2(0.6)^{ab}

Table 18: Significance results comparing mean hysteresis values between bracket systems within a wire material

Wire Material	Significance Results
NiTi	$F(9,90) = 55.77$, $P<0.001$
TMA	$F(9,90) = 18.92$, $P<0.001$
SS	$F(9,90) = 4.63$, $P<0.001$

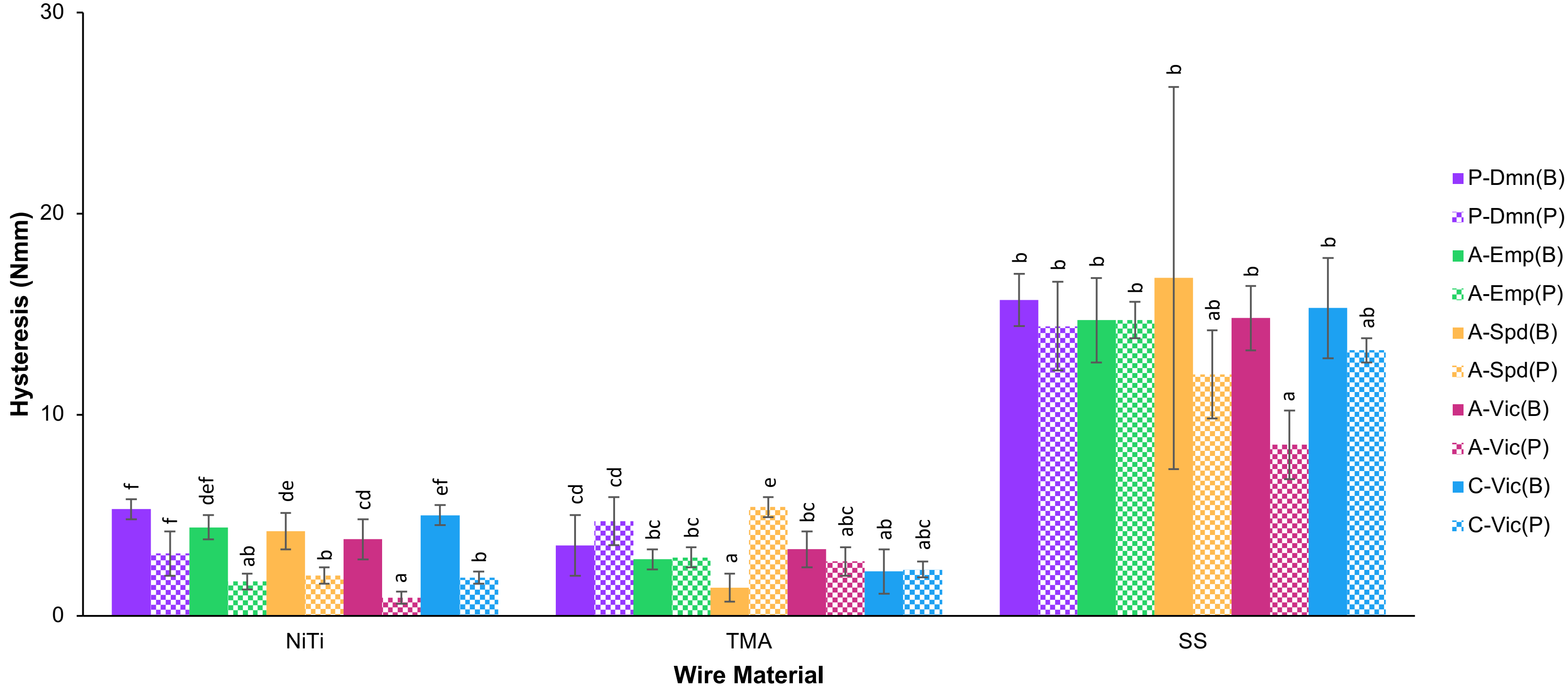

Figure 36: Mean hysteresis values in Nmm (±SD) of different bracket systems versus wire material. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05

Table 19: Mean hysteresis values in Nmm (±SD) for each bracket system-wire combination. Non-significant differences between wire materials within a given bracket group at P>0.05 are denoted by shared alphabetical letters within each row

	Wire Material		
Bracket System	NiTi	TMA	SS
P-Dmn(B)	5.3(0.5)[b]	3.5(1.5)[a]	15.7(1.3)[c]
P-Dmn(P)	3.1(1.1)[a]	4.7(1.2)[a]	14.4(2.2)[b]
A-Emp(B)	4.4(0.6)[b]	2.8(0.5)[a]	14.7(2.1)[c]
A-Emp(P)	1.7(0.4)[a]	2.9(0.5)[b]	14.7(0.9)[c]
A-Spd(B)	4.2(0.9)[a]	1.4(0.7)[a]	16.8(9.5)[b]
A-Spd(P)	2.0(0.4)[a]	5.4(0.5)[b]	12.0(2.2)[c]
A-Vic(B)	3.8(1.0)[a]	3.3(0.9)[a]	14.8(1.6)[b]
A-Vic(P)	0.9(0.3)[a]	2.7(0.7)[b]	8.5(1.7)[c]
C-Vic(B)	5.0(0.5)[b]	2.2(1.1)[a]	15.3(2.5)[c]
C-Vic(P)	1.9(0.3)[a]	2.3(0.4)[a]	13.2(0.6)[b]

Table 20: Significance results comparing mean hysteresis values between wire materials within bracket systems

Bracket Group	Significance Results
P-Dmn(B)	F(2,27) = 319.82, P<0.001
P-Dmn(P)	F(2,27) = 156.57, P<0.001
A-Emp(B)	F(2,27) = 252.03, P<0.001
A-Emp(P)	F(2,27) = 1293.67, P<0.001
A-Spd(B)	F(2,27) = 22.05, P<0.001
A-Spd(P)	F(2,27) = 141.12, P<0.001
A-Vic(B)	F(2,27) = 307.38, P<0.001
A-Vic(P)	F(2,27) = 143.77, P<0.001
C-Vic(B)	F(2,27) = 186.77, P<0.001
C-Vic(P)	F(2,27) = 2077.18, P<0.001

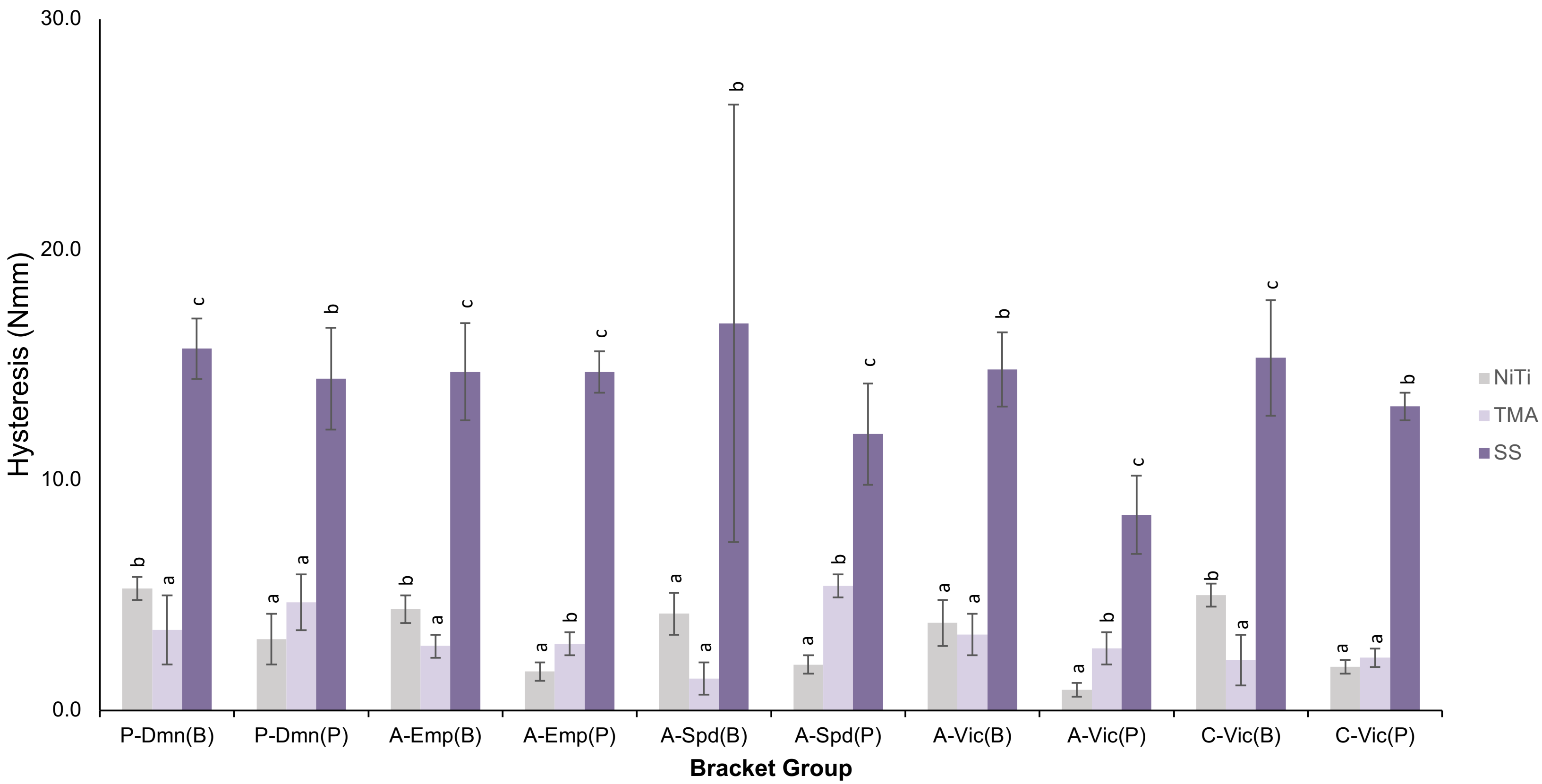

Figure 37: Mean hysteresis values (±SD) with different wire materials versus bracket systems. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wire materials at P>0.05

Chapter 5

5 Discussion

The primary aim of this investigation was to evaluate torquing moments generated by ASL orthodontic brackets when compared with a PSL system and a conventionally ligated control with various archwire materials in both buccal and palatal root torque simulations. This investigation set out to build upon a previous study investigating torque moments generated with various bracket systems and archwire sizes[68], and thus intended to explore the reported benefits of ASL systems, particularly the suggestion that the wire-seating mechanism of these brackets leads to the generation of larger torquing moments. This investigation also set out to examine the relationship between engagement angle, torsional stiffness, and hysteresis when comparing various archwire materials and ligation methods of the brackets of interest. In contrast to previous studies of orthodontic torque expression, this investigation was unique in that it explored torque expression of the most current and commonly used bracket systems on the market (ASL, PSL and conventionally ligated) while using numerous wire materials in two directions of rotation (simulations in buccal and palatal root torque). A custom table-top measuring apparatus fabricated using 3D printing technologies was developed to undertake the investigation, and the results would be expected to allow clinicians to better understand the functionality of the orthodontic appliances they utilize, and ultimately facilitate customization of orthodontic appliances based on the torquing needs of an individual patient.

5.1 Study Methodology

The current investigation used a newly developed, custom fabricated torque assembly to study torque in orthodontics, which has some unique features while still maintaining similarities to other modalities of studying torque that have been reported in the literature. The setup used in this study was similar to the apparatus developed by Badawi et al[66] which used a digital inclinometer to measure the degree of torsional rotation, a wire support substructure to hold the wire, and a worm-gear to rotate the wire segment. The current setup is also similar to the apparatus used a previous thesis, which used an Instron materials testing machine to measure torquing moments generated with a fixed segment of wire.[67] Both of these similar testing devices used wire spans of 15 mm, as did the current investigation, which is important in comparing results given the known relationship between wire length and stiffness.[1]

The current investigation's methodology differed from these previous studies in that the wire of interest was held stationary while the bracket was rotated around the wire, in contrast to the bracket being held stationary while torsion was applied to the wire segment. Mechanically, the same process occurs within the bracket slot in both situations, such that one would expect the findings to be similar. The current investigation utilized visual inspection to ensure proper alignment of the bracket slot and wire, which was consistent to the methods used by Young[67], but contrary to those used by Badawi et al[66] who used turntables to adjust the bracket position to ensure all other forces were zeroed. Regardless, the mean torquing moments and standard deviations from the current investigation are comparable to those reported by Badawi, which lends support to the effectiveness of the visual alignment approach used. The use of the small capacity load cell in this investigation should be considered an advantage which provided improved precision and accuracy, especially in comparison to Young.[67] Their apparatus used a 10 kilonewton load cell, which has been shown to produce significant measurement noise, and this may limit the accuracy of results at low angles of torsion and small torque values.

The current apparatus using a fixed segment of wire should not be directly compared to those studies who utilized the OMSS system for torque testing. Seeing that the OMSS models a complete dental arch and uses a continuous archwire, it would be expected that torque moments measured would be lower and engagement angles would be higher than this investigation, due to the adjacent teeth in the model providing increased archwire play.[22]

The analysis of torque moments in both buccal and palatal directions provides insight into the behaviors of the given bracket systems, which many of the previous studies on torque expression in orthodontic brackets lack. Few studies were found which have also studied this effect on ligation method and torque expression.[31,67,68] In clinical practice both directions of torque may be applied to a tooth, and given that is has been shown the direction of rotation can have significant influences on torque expression generated by some bracket systems, it was important to study the influence of both directions when examining different archwire materials. This allows one to be able to judge the full range of a given bracket's capabilities of torque expression.

The slight modifications made to the current apparatus in comparison to the investigation by Boogaards[68] were thought to provide improved precision during testing. It was noted that the alignment jig previously used to establish the repeatable zero position allowed for a slight degree of play. It was felt that the addition of the limit switch would eliminate this potential issue, as it was now possible to program the device to return to a computed zero position rather than relying on the alignment jig. It was also noted that the literature was lacking on torquing moment data which would occur on the unloading curve, with only a few studies available that have studied SL brackets.[34,65] The programming of the current apparatus was adjusted to allow the device to first complete a loading cycle, pause, and then complete an unloading cycle, all while collecting data, to add to the literature on unloading curves of SL brackets.

5.2 Mean Torquing Moments with Progressive Rotation

In examining torque moments produced with progressive rotation of a bracket around a wire of interest, regardless of the bracket system, direction of rotation (buccal or palatal root torque simulations), or wire material, mean torquing moments increased with increasing twist after engagement with the archwire within the bracket slot. This is highlighted in Figures 14-16, and is consistent with the available literature on the subject.[24,31,34,42,44,65,66]

In direct comparison to similar available literature (comparable methodology, bracket system, wire material and size, and degree of rotation), torquing moments generated in the current investigation are comparable to those previously recorded. For example, when comparing the loading curves of SS wires, Major et al[34] found the Damon system generated a mean torquing moment of 9.6 Nmm, 50.1 Nmm, and 86.9 Nmm at 15, 30, and 45 degrees of rotation respectively. The torquing moments generated during this investigation with P-Dmn(B) were found to be 18.0 Nmm, 52.2 Nmm, and 84.8 Nmm at 15, 30, and 45 degrees respectively. Given that Major et al[34] reported standard deviations as high as 5.5 Nmm within these degrees of rotation with the Damon system, one can consider these values comparable. In comparing to the previous thesis by Boogaards[68], whose methodology most closely matches that of the current study, and examining torquing moments generated by the Empower Interactive brackets, a mean moment of 9.6 Nmm, 38.2 Nmm, and 70.1 Nmm were generated at 15, 30, and 45 degrees of rotation respectively when examining a 0.019 x 0.025-in SS wire in a buccal root torque simulation. The current investigation found mean torquing moments generated in the A-Emp(B)/SS group to be 8.2 Nmm, 39.3 Nmm, and 72.4 Nmm at 15, 30, and 45 degrees respectively. When considering standard deviations, these values are found to be comparable.

In examining the torque-rotation curves of TMA and SS wires, the relationship between torque and degree of rotation was generally linear after engagement of the wire within the slot, with the exception of the A-Spd/SS group in the buccal root torque direction. In examining the A-Spd(B)/SS group, there is a loss of linearity of the curve as the rotation exceeds 40 degrees, resulting in a decline in torquing moments. This pattern has been observed in previous studies.[34,42,66,68] A possible explanation for this is that the clip mechanism used by the A-Spd brackets is made of NiTi, and it is possible that had the rotation progressed further the curve would have followed the expected force-deflection curve characteristic of NiTi material. At this level of rotation and force applied to the NiTi clip, it is possible it is undergoing the stress-induced phase transformation from A-NiTi to M-NiTi.[75] During testing it was observed that the clip on the A-Spd brackets was deforming outward such that it was unable to actively hold the archwire into the bracket slot any further. As the bracket progressed back along the unloading curve, the clip returned to its original position, and torquing moments observed were linear. This indicates that the deformation observed of the clip was elastic in nature. It is also possible that there was some plastic deformation that was occurring in the system which resulted in the decline in torquing moments. It is unlikely that this is a major factor in the differences only seen with the A-Spd brackets, as shown by the hysteresis noted between loading and unloading curves of all bracket systems tested (see Table 17).

In examining the torque-rotation curves for NiTi wires it was noted that beyond approximately 30 degrees of rotation, the linear trend between torque and degree of rotation was not maintained, and the curve began to reduce in slope at high degrees of rotation. This pattern has been previously demonstrated in the literature.[73,75,76] This is seen due to the superelastic nature of NiTi wires, in which A-NiTi is undergoing a stress induced transformation to M-NiTi. Here the wires do not follow Hooke's law, and exert the same amount of force independent of the degree of activation. In a study on NiTi wires of smaller dimensions, Meling and Ødegaard[73] found that this plateau occurred beyond 25 degrees of rotation. Bolender et al[75] also studied torque-rotation curves of 0.017 x 0.025-in NiTi wires in 0.018-in bracket slots and found similar results, in which wires demonstrated a horizontal plateau between 20 to 25 degrees of rotation. These finding are consistent with the plateau noted in the torque-rotation curves for NiTi wires in this investigation.

5.2.1 Comparing Bracket Systems

Comparing mean torques generated by the various bracket systems of interest revealed that at 15 degrees of rotation, many non-significant differences between systems were noted. This observation of generally no difference between bracket systems at low degrees of rotation has been demonstrated in the previous literature. Major et al[34] noted that generally below 25 degrees of rotation one would not find significant differences between ASL and PSL bracket systems in studying 0.019 x 0.025-in SS wires. While this investigation did show that at 15 degrees of rotation there does exist some significant differences between bracket systems, it remains to be seen if these differences are clinically significant, especially with NiTi wires. Given that a torque moment of 5 Nmm is considered the threshold for tooth movement [22–24], it is unlikely that any bracket group would be capable of producing torque at 15 degrees of rotation with NiTi wires. Only the P-Dmn(B) group produced a torquing moment above 5 Nmm but this was not significantly different than the C-Vic(B) system which produced a mean torquing moment of less than 5 Nmm.

Interestingly, as rotation progressed through to higher values with the NiTi wires there were fewer significant differences noted between bracket groups. It is likely that this finding is more directly related to the properties of the NiTi wires. In their study on torsion of NiTi wires, Partowi et al[76] found that these wires do indeed exhibit a superelastic plateau when subjected to torsion. This plateau began at approximately 20 degrees for wires of small diameters (0.016 x 0.022-in) and approximately 30 degrees for wires of larger diameters (0.019 x 0.025-in) which aligns well with the findings of this investigation. Therefore, it is apparent that when considering torque expression with NiTi wires, one can expect similar expression regardless of the method of ligation, with all bracket systems being capable of producing clinically relevant torquing moments at or above 30 degrees of twist (see Table 3).

For all degrees of rotation for both TMA and SS wires, generally P-Dmn and C-Vic systems tend to produce significantly greater torquing moments than the ASL systems. At 15 degrees only P-Dmn and C-Vic produced torquing moments which would be considered clinically relevant (above 5 Nmm) in both directions of rotation. With the ASL systems at 15 degrees, only rotation in the buccal root torque direction resulted in clinically relevant torque moments. At 30 degrees of rotation, all bracket systems met or exceeded the clinically relevant range, while at 45 degrees all bracket systems exceeded 20 Nmm of torque. This pattern has also been observed in the literature. Huang et al[77] found that PSL and conventionally ligated brackets produced greater torquing moments when compared to an ASL system, up to rotations of 20 degrees. A previous thesis also found similar comparisons for the same selection of brackets.[68] Yet this is in contrast to a systematic review which concluded that conventionally ligated bracket systems demonstrated greater torquing moments when compared with SL systems, and only minor differences were noted between ASL and PSL systems.[24] It is likely that the conflicting results are at least partially due to the differing methodologies of studying torque expression in orthodontic brackets that were included in the systematic review.

It was hypothesized that the ASL bracket systems would produce greater torquing moments, however this was not found in the current investigation. It is possible that other aspects of the bracket design have a greater impact on torque expression than the ligation method. A study comparing torque expression of ASL brackets with doors open and closed by Brauchli et al[31] found that only 1 Nmm of additional torque moment was observed when the ASL clips were closed compared to when the clips were opened. While this study only rotated test brackets to a maximum of 30 degrees, mean torquing moments observed were in line with those of the current investigation. Specifically, Brauchli noted 15.9(±1) Nmm of torque expression in Speed brackets when rotated in a palatal root torque direction with the clip closed. The torque expressed by the A-Spd(P) group at 30 degrees in this investigation was 18.8(±3.1) Nmm. Taking into account the SDs, one can conclude these results are in agreement with one another, adding to the validity of the current investigation. Brauchli concluded that both bracket material and slot dimensions play a larger role in torque expression than does ligation method. Indeed, a study by Cash et al[78] completed in 2004 cautioned clinicians on the loss of torque control of teeth due to bracket slot dimensions being larger than stated by the manufacturers. It was shown that the brackets could vary as much as 24% larger than the dimensions provided by the manufacturer. It has also been shown that bracket material can have an impact on torque expression. Major et al[79] demonstrated that plastic deformation of the bracket slot occurs with torque expression, in both PSL and ASL brackets. They found that the ASL system demonstrated greater plastic deformation, which also occurred at earlier degrees of rotation, when compared to the PSL system. These findings align with the results of this investigation.

5.2.2 Comparing Wire Materials

When comparing torquing moments generated by the various bracket systems with different wire materials, it was shown that SS wires tended to generate significantly larger torquing moments than the TMA and NiTi wires, regardless of the degree of rotation. Intermediately, TMA produced mid-range torquing moments when compared with SS and NiTi. The lowest torquing moments were generated with NiTi wires. These results are not unexpected, given that SS is the stiffest of the three materials, followed by TMA, and then NiTi.[17] This observation has been seen in the literature, with Archambault et al[42] also noting torquing moments were greatest with SS wires when compared with TMA and NiTi in SL bracket systems. They noted SS wires produced approximately 1.5

Table 21: Mean torquing moments presented as ratios of SS for each bracket-wire combination for every 15 degree increment of rotation on the loading curve

		Wire Material		
Bracket System	Degrees	NiTi	TMA	SS
P-Dmn(B)	15	0.3	0.6	1.0
	30	0.3	0.7	1.0
	45	0.2	0.7	1.0
P-Dmn(P)	15	0.2	0.9	1.0
	30	0.3	0.8	1.0
	45	0.2	0.7	1.0
A-Emp(B)	15	0.5	1.0	1.0
	30	0.4	0.8	1.0
	45	0.3	0.7	1.0
A-Emp(P)	15	0.3	1.1	1.0
	30	0.4	0.8	1.0
	45	0.4	0.8	1.0
A-Spd(B)	15	0.6	0.9	1.0
	30	0.4	0.7	1.0
	45	0.3	0.8	1.0
A-Spd(P)	15	0.3	1.3	1.0
	30	0.4	0.8	1.0
	45	0.4	0.9	1.0
A-Vic(B)	15	0.3	0.9	1.0
	30	0.3	0.8	1.0
	45	0.3	0.7	1.0
A-Vic(P)	15	0.9	2.5	1.0
	30	0.4	1.1	1.0
	45	0.4	0.9	1.0
C-Vic(B)	15	0.5	0.7	1.0
	30	0.4	0.7	1.0
	45	0.3	0.7	1.0
C-Vic(P)	15	0.3	0.8	1.0
	30	0.3	0.7	1.0
	45	0.3	0.7	1.0

to 2 times greater torquing moments than TMA wires at angles above 24 degrees. This finding was slightly higher than the results of the current investigation (see Table 21). In comparing SS wires to NiTi, Archambault found the torque expression to be 2.5 to 3 times higher in the SS groups. This investigation noted torque expression in SS ranged from 3 to 4 times higher than NiTi, with bracket groups on the higher end of the range being the PSL and conventionally ligated systems. While these comparisons between SS and NiTi wires do not align with Archambault, they more closely align with the findings from Meling and Ødegaard[73] in which they reported a ratio of the modulus of elasticity in tension to be 3.7 between SS and NiTi.

The most apparent reason for the differences in torque moments is related to the modulus of elasticity of the wires tested. Given that SS has the highest modulus of elasticity, it was expected that it would generate the largest torquing moments, followed by TMA intermediately, and that NiTi wires with the lowest modulus of elasticity would generate the lowest torquing moments. As a result of the reduced modulus of elasticity of TMA and NiTi wires in comparison to SS, it has been stated that wires of reduced modulus are ineffective at transmitting torquing moments to bracket slots.[41] This investigation found contradictory evidence to this statement, and found that both NiTi and TMA wires are effective at producing torquing moments in all bracket groups, regardless of ligation method. While some of the bracket groups were unable to reach the upper end of the biologically acceptable torquing moment range of 20 Nmm, all still produced torquing moments greater than the low range of 5 Nmm, thus refuting the above statement.

5.3 Torquing Direction

When comparing mean torquing moments generated by bracket systems in simulations of buccal and palatal root torque, it was noted that the direction of rotation tended to influence the torquing moments generated by the ASL systems but not those of the PSL or conventionally ligated systems. Palatal root torque tended to produce significantly lower torquing moments than buccal root torque simulations with the ASL systems. This finding aligns with those reported in other studies. In studying Speed brackets, Brauchli et al[31] also observed significantly lower torquing moments when palatal root torque was simulated when compared to buccal root torque. Similarly, a previous thesis found comparable results, in which torque moments generated through palatal root torque simulations were significantly lower than those generated through buccal root torque simulations for ASL brackets, but not for conventionally ligated or PSL bracket systems.[68]

These findings can in part be explained by the bracket slot design, and through which it allows the active clip mechanism to seat the archwire in ASL systems. In examining these bracket slots, one can appreciate that the gingival portion of the slot is shallower than the incisal portion. As an archwire twists within the slot to produce palatal root torque (Figure 38A), it is possible that the wire could deform the door outward and slip past the gingival portion of the slot. This would result in a loss of torquing moment, as the wire is unable to fully engage both walls of the slot to create the couple needed to produce torque. Conversely, in the buccal root torque direction this loss in torquing moments would not occur, since the wire is rotating in the opposite direction such that it will not slip past the shallow slot depth of the gingival wall (Figure 38B). All of the slot depths of the ASL brackets tested in this study are less than that of the depth of the wires investigated (0.025-in). The minimum slot depths for the ASL systems investigated were found to measure 0.0189-in, 0.0153-in, and 0.014-in for A-Vic, A-Spd, and A-Emp, respectively.[72]

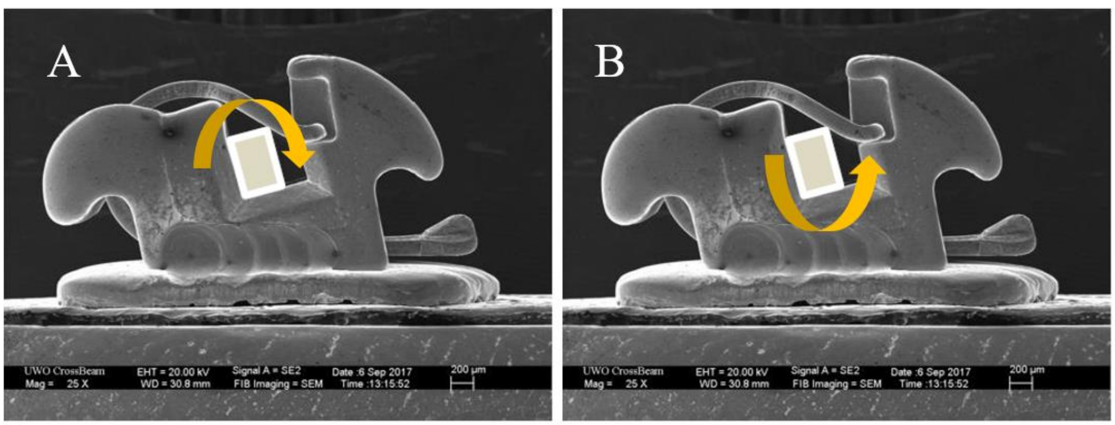

Figure 38: Scanning electron microscopy view of a self-ligating bracket, demonstrating the discrepancy in slot depth between incisal and gingival aspects to allow for seating of the active clip. (A) demonstrates the wire twist required to generate palatal root torque, (B) demonstrates the wire required to generate buccal root torque. Image used with permission from previous thesis completed by Greene[72]

Another explanation for the difference in torquing moments relates again to the difference in the depth of the bracket slot walls. In studying the deformation of brackets due to torquing forces, Major et al[61] found that the gingival wall of Speed brackets showed substantial deformation. This was due to the relatively smaller amount of bracket material to support it when compared with the incisal wall, because of the position of the groove for the bracket clip.

This finding of significantly different mean torquing moments between buccal and palatal root torque simulations with ASL, but not with PSL or conventionally ligated systems, has implications when comparing the results of this investigation to other studies, as well as in applying the findings to clinical scenarios. Based on whether previous studies have examined buccal or palatal root torque simulations could have a connection to their conclusions regarding the influence of ligation method on torque expression. For example, if a previous study only focused on buccal root torque simulations, ASL systems would appear to have relatively superior performance compared to other ligation methods. If only palatal root torque simulations were examined, ASL systems would demonstrate relatively poorer performance. In addition, since every action has an equal and opposite reaction, it could be predicted that an average of the buccal and palatal root torque moments would give a more accurate representation of the overall torque values one could expect to produce in a clinical scenario with each bracket system. Thus, studying torquing direction was an important component of the current investigation.

5.4 Engagement Angles

In comparing engagement angles with different wire materials within bracket groups, it was found that, in general, significantly larger mean engagement angles were noted with NiTi wires, while the smallest engagement angles were noted for the stiffer TMA and SS wires. However, this was not always the case, as in the A-Spd(B) and C-Vic(B) groups in which there was no significant difference between the three wire materials. Given that all the wires were the same dimension, and it has been observed that wire size has an impact on engagement angle, this result of the lack of significant difference between wire materials was expected.[33] In the remaining bracket groups were there were differences noted between wire materials. In most groups the TMA wire demonstrated the smallest engagement angle, while in P-Dmn(B) group both TMA and SS wires were found the produce the smallest engagement angles. To date, there does not appear to have been any other investigation completed on engagement angle differences between wire materials, and as such there is no data to compare to. However, several possible explanations exist as to why, in most bracket groups, the TMA wire produced the smallest engagement angles.

The first relates to the stiffness of the wire materials. It is possible that due to its increased rigidity, the SS wire produced some plastic deformation of the bracket slot prior to reaching its engagement angle. Several studies have shown that SS wires do indeed deform bracket slots in torque expression [61,79,80], however Major et al found that this deformation occurs at approximately 28 and 24 degrees of rotation for Damon and Speed bracket, respectively. The second possible explanation is the frictional differences between wire materials. TMA is known to have a high coefficient of friction[17], therefore it is possible this plays a role in the wire engaging the slot prior to SS. The last explanation could have to do with manufacturing tolerances of the archwire dimensions and bevel. It has been well demonstrated that there exist differences between reported values and measured values for both wire height and width, as well as the wire bevel. [35,38,73,74] Of the wires utilized for this investigation it was found that the TMA wires had a smaller edge bevel than the NiTi or SS wires (Appendices G-I), which is likely the main factor that allowed TMA wires to engage in the bracket slots earlier. This finding was contradictory to previous studies on edge bevel, which have shown that TMA wires tend to have the largest edge bevel, and edge bevels between SS and NiTi are not significantly different.[81,82] Further research on this topic should be undertaken to better understand these findings, as this investigation only considered wires from one manufacturer. Differences between manufacturers exist, and as such a better understanding of the amount of edge beveling that each manufacturer tolerates would be advantageous when determining what torque moments and engagement angles could be anticipated clinically.

When comparing engagement angles between bracket groups, few patterns emerged across the wire materials. Generally the P-Dmn brackets tended to produce the smallest engagement angles across all wire materials, but this observation was only significant in the SS wire group. With regards to direction of rotation and engagement angle, it was found that for all bracket systems there were significant differences between buccal and palatal root torque simulations, with the exception of the A-Spd and C-Vic groups in both

TMA and SS wires. This finding relates back to bracket design intricacies, such as slot dimensions and ligation methods.

5.5 Torsional Stiffness

Comparing the torsional stiffness of the archwire materials within bracket groups, it was shown that SS produced the largest torsional stiffness values, followed by TMA intermediately, and NiTi produced the smallest values. These findings were significant for all bracket groups. The values observed in the current investigation (Table 15) align with those completed by Meling et al[32,73], where they observed approximately 4 Nmm/° for SS wires, 1.5 Nmm/° for TMA wires, and 0.8 Nmm/° for NiTi wires. These results are not surprising, given what is known about the modulus of elasticity of the given wires.

In comparing the torsional stiffness values of the bracket groups within wire materials, little variation was noted for the more flexible wires of NiTi and TMA. In examining the SS group, generally P-Dmn and C-Vic demonstrated the highest values, while the ASL systems showed lower values. As this measure of torsional stiffness is a measure of the entire system (bracket + archwire) it can be hypothesized that the PSL and conventionally ligated bracket are more rigid than the ASL systems, which led to the higher torsional stiffness values. In considering that the Speed system uses a NiTi clip, this would reduce the rigidity of the system compared to the rigid door of the Damon system. Several studies conducted by Major et al[61,79,80] have shown that there is indeed more plastic deformation that occurs in Speed brackets compared to Damon brackets. They have hypothesized that due to the groove in the gingival wall of the Speed bracket slot which houses the clip, there is less material to support this wall, decreasing its strength, which has led to the observations of higher plastic deformation in Speed brackets. Indeed, the ASL systems investigated in this current study demonstrate differences in the dimensions of the occlusal and gingival walls of the slots, whereas with the PSL and conventionally ligated systems the walls are the same dimensions or only slightly different (see Figure 7). It is possible that this difference in wall height between bracket groups plays a role in the torsional stiffness of the system as a whole.

5.6 Hysteresis

All hysteresis values were found to be positive, regardless of bracket system or wire material, demonstrating that torque expression is lower on the unloading curves when compared with the loading curves. From a clinical standpoint, the unloading curve is arguably more important than the loading curve, because when a twisted wire is inserted into a bracket, the loading action occurs near instantly, while the unloading action is sustained throughout active tooth movement.[65] In their study on SS wires in SL brackets, Major et al[34] felt that the plastic deformation of the wire accounted for most of the difference, while the plastic deformation of the bracket, although not as substantial as wire deformation, could additionally reduce the torque expression while the system is unloading. Fakir et al[65] also found similar results in differences between loading and unloading curves of SS wires and SL brackets, and was in agreement with Major in that the change is likely a result of plastic deformation of the wire and/or bracket.

Generally, it was found that regardless of bracket system SS wires significantly produced the highest levels of hysteresis. As the highest torquing moments generated in this investigation were with SS wires, and it is the stiffest wire material tested, it is likely that the most plastic deformation occurred with these groups. Thus, logically one would expect to find the highest hysteresis in these groups. Between the NiTi and TMA wires, it was found that NiTi produced the lowest hysteresis values in palatal root torque directions, while TMA produced the lowest values in buccal root torque directions, for all bracket systems, regardless of ligation method. The finding of NiTi wires producing higher hysteresis in the buccal root torque direction is related to the finding that higher torquing moments were found in this direction of torque, and also because of the superelastic characteristic of the wire, in that the reversibility between A-NiTi and M-NiTi is associated with energy loss.[1]

5.7 Clinical Recommendations

As previously discussed, torquing moments for biologically acceptable tooth movements are cited to range from 5 to 20 Nmm. [22–24] It would be desirable to determine the minimum and maximum degrees of twist a clinician would be required to place into a given wire material to generate torquing moments within this range. These values were calculated for the bracket systems and wires used in this investigation and can be found in Table 22.

All the wire material-bracket combinations were capable of generating, at minimum, 5 Nmm of torque within the 45 degrees of rotation simulated in this study. The TMA and SS groups were all capable of producing torque within the biologically accepted range. All of the buccal root torque NiTi groups were capable of producing torque between 5 and 20 Nmm, with the exception of the A-Vic(B) group which maxed out at 19.3 Nmm of torque at 45 degrees. None of the palatal root torque NiTi groups were able to reach 20 Nmm of torque expression within the 45 degrees of rotation. These findings contradict that of Morina et al[41] who claimed that because NiTi and TMA wires present with only a fraction of the torsional stiffness of SS, and have reduced hardness, they would be incapable of transmitting torque moments to bracket slots.

With progressively larger amounts of rotation and the stiffer wire materials of TMA and SS, torquing moments much larger than the biological threshold were generated. For example, the largest torquing moment of this investigation was recorded at 45 degrees of rotation with SS wires in the P-Dmn(B) group, a mean value of 84.8 Nmm (Table 4). Practically speaking, one may not expect to be able to produce such a torquing moment clinically, due to the fact that with large dimension wires, and large degrees of twist, it becomes increasing difficult to engage the wire within the bracket slot, and/or close the bracket door. As a result, some of the simulations of the investigation are not necessarily representative of a clinical situation.

Generally, as the wire stiffness increased, the amount of twist required to generate biologically acceptable torque ranges decreased. However, regardless of the wire material, the degree of twist required to generate a clinically significant torque moment was often greater than many common bracket prescriptions (+7 for Andrews, +12 for Roth, +17 for MBT), which casts doubt on the clinical relevance of different bracket prescriptions with the wire materials investigated in this study. This finding is not unique, as others have noted that torque play may be enough to cancel out incorporated torque prescriptions.[31,41,83] Clinical investigations have supported this, reporting that bracket prescriptions have no clinical influence on treatment outcomes[84], and no effect post-treatment outcomes of subjective esthetics.

Of note to clinicians should be the range of twist required to generate biologically acceptable torquing moments, which was calculated from data collected during the current investigation and can be found in Table 23. It can be seen that with increasing wire stiffness, the range of rotation required to produce 5 to 20 Nmm of torque decreases. Based on this finding, one must exercise caution when placing twist into a SS wire, given the narrow working range of biologically acceptable torque.

The long range of the NiTi wires proves advantageous, in that even at high degrees of twist the torquing moments produced still fall within the biologically acceptable range. This is due to the superelastic plateau exhibited by the NiTi material, and acts as a safeguard to producing torquing moments outside of this range, In addition, NiTi wires store larger amounts of energy, so that the patient does not need to be seen by the clinician as often for wire reactivations or changes. These are ideal characteristics of a wire when attempting to complete orthodontic treatment with light, continuous forces. The drawback to torquing with NiTi is the lack of formability of the wire, making it difficult to place torquing bends if additional torque beyond the bracket prescription is required. Pre-formed torquing wires are available on the market today, for example Ormco produces a 20 degree pre-torqued NiTi archwire. If this wire was used in a MBT prescription of +17, it could effectively produce a torquing moment equivalent to 37 degrees of twist, which would be within the ideal range of 5 to 20 Nmm for all brackets investigated in this study.

The TMA wire appears to be the “Goldilocks” option for archwire of choice when producing torquing moments. It produces a biologically acceptable range over a larger degree of wire twist when compared with SS. It has better formability than NiTi, such that placing torquing bends is simple when necessary. In addition, this investigation found that it produces the lowest engagement angles, allowing for torque expression to occur earlier in comparison to SS and NiTi wires.

Given the experimental design of the investigation, these ranges are likely an underestimation of what one would expect clinically. This is due to the fact that the wires were firmly clamped on either side of the test brackets, rather than being held in an adjacent bracket as they would be in a clinical scenario. With brackets on either side of the bracket of interest, the torque play in the adjacent brackets will add to the play in the system, which subsequently would increase the degree of twist needed to generate a clinically relevant torquing moment. Nonetheless, clinicians can consider these ranges as a rough guide.

Table 22: Degree of rotation required to generate mean torquing moments of 5 and 20 Nmm for each bracket-wire combination examined. N/A indicated 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)	
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 Material	NiTi	13.7	43.7	18.1	N/A	16.9	40.7	24.1	N/A	15.6	40.7	25.3	N/A	16.8	N/A	27.7	N/A	15.1	40.3	20.5	N/A
	TMA	9.1	21.5	8.0	19.9	11.6	24.0	17.3	31.4	12.7	25.4	15.4	35.0	10.0	22.6	17.3	31.8	13.3	24.5	13.5	25.4
	SS	7.4	16.0	9.3	18.2	12.8	21.5	16.9	27.5	12.9	21.4	16.4	31.4	10.5	19.6	20.7	32.3	11.9	20.4	13.3	21.7

Table 23: 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 5 Nmm. N/A indicated the threshold mean 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 Material	NiTi	30	N/A	23.8	N/A	25.1	N/A	N/A	N/A	25.2	N/A
	TMA	12.4	11.9	12.4	14.1	12.7	19.6	12.6	14.5	11.3	11.9
	SS	8.6	8.9	8.7	10.6	8.4	15.0	9.1	11.6	8.4	8.4

5.8 Strengths and Limitations of the Study

The first advantage of the current investigation is the development of the novel torque-testing apparatus. Due to its affordable and portable 3D printed table-top design, it could allow numerous others to utilize the same design repeatedly, which is paramount for the future of research. In addition, the use of the small capacity load cell improved accuracy over previous study designs which utilized larger load cells, by reducing measurement noise.[67] Regarding the methodology of the investigation, determining the engagement angles through the use of linearity equations offers an advantage over some previous research which determined engagement angles by reading off of graphs.[66,67] In addition, the collection of both the loading and unloading torquing moments is under reported in the literature, and can offer insights into the force system felt by teeth once a torquing moment is applied.

This investigation examined all three commonly used ligation methods in current practice, including multiple commonly available and utilized ASL systems, in addition to the three most commonly used wire material alloys. In addition, torque expression was examined in both buccal and palatal root torque directions. This range of comparisons allowed for the examination of factors influencing torque expression that have been only marginally considered in the literature up to this point. This allows for comparisons obtained from the current investigation to be considered more valid, as they have all been examined under the same methodology, in contrast to attempting to compare factors across numerous studies with varying methodologies.

One weakness of the current methodology is the *in vitro* approach and the usefulness of applying the findings to clinical practice. Wires of interest were held on either side of the brackets in a rigid clamp, to ensure repeatability of the investigation. This does not reflect an *in vivo* scenario, in which the wire is held by other brackets on either side of the bracket/wire segment of interest. Brackets *in vivo* are bonded to teeth, which are surrounded by both hard and soft tissues, such as bone and the PDL. The current study design was not able to take this into account. As a result, the findings of this investigation are likely higher than what one would expect to find in a clinical scenario. Additionally, the investigation was completed in air at room temperature whereas clinically these torquing moments are applied within the mouth, which is a moist environment at body temperature. It has been shown that elastomeric ligature force decay and deformation are affected by moisture and heat, such that conventionally ligated systems may perform differently clinically.[64] Also, NiTi material properties change with temperature[17], thus both the NiTi archwires and the NiTi clip of the Speed brackets may perform differently in an *in vivo* situation. Lastly, it has been discussed that plastic deformation can occur with a single application of torque.[61,65,79,80] Considering that orthodontic treatment requires on average 30.1 months for completion in adolescents[85], and the brackets remain in place for the duration of treatment, it is reasonable to expect there to be increased play in the bracket as treatment progresses. As well, fatigue of the clips and doors of the SL system may also be a factor leading to increased play in the system throughout the treatment period.

As previously noted, slight misalignments between the bracket slot and wires of interest due to confirmation of alignment through visualization may present another potential limitation of the study design. While Romanyk et al[86] did find that second-order misalignment produced significant differences in torque expression, the magnitude of the differences was likely not clinically significant considering the range of torquing moments required for biological tooth movement, variation in biological tissues in patients, and appliance tolerances. Considering the observed consistency in standard deviations across most bracket groups for both torquing moments and engagement angles, it is not expected that this limitation was significant across the investigation.

5.9 Future Research

There are ample opportunities for further research in the field of orthodontic torque expression. Future development of a torque-testing apparatus that could be used in a micro computed tomography scanner could provide insights into the wire-bracket interactions during torque application. This could help to reveal why certain patterns were observed in this study, such as the directionality of torque expression with ASL brackets, the smaller engagements angles of TMA wires and PSL brackets, and the differences in torsional stiffness hysteresis between bracket systems. In addition, this could provide insight into real-time distortions of both the bracket and archwire as they are undergoing torque application.

A more in depth investigation on archwire dimensions would also prove helpful in determining why certain patterns were observed in this study, specifically related to edge bevel but more generally as well. Comparing archwires from different manufacturers, and archwire materials in various sizes could uncover further differences in torque expression. Additionally, torque expression in 0.018-in bracket slots with various archwire materials would provide further information for clinicians to be able to make informed decisions on the choice of appliances they utilize.

Attempting to design an investigation that more closely replicates clinical scenarios would also prove insightful for clinicians. Ideally such an investigation would be completed with repeated torque cycles with one bracket and wire segment to demonstrate degradation and fatigue over time. This investigation would also be completed in a warm, moist environment that more closely mimics the oral environment.

Chapter 6

6 Conclusions

An *in vitro* examination of torquing moments generated with three ASL, a single PSL, and a single conventionally ligated bracket using various archwire materials revealed:

1. Greater torquing moments were produced with the PSL and conventionally ligated bracket systems, when compared to the ASL brackets, especially with stiffer wire materials and at greater degrees of rotation.

2. Stiffer wire materials produced greater torquing moments than more flexible wires, regardless of bracket system or degree of rotation applied.

3. Direction of torque influenced torquing moments generated with ASL systems but not PSL or conventionally ligated systems, with buccal root torque simulations producing significantly greater torquing moments than palatal root torque simulations.

4. Engagement angles were lowest for TMA wires, followed by SS, and highest for NiTi wires. Between bracket systems, the PSL bracket tended to produce the lowest engagement angles.

5. Hysteresis values were highest with SS wires, and lowest with NiTi wire, regardless of bracket system or direction of rotation.

6. Ligation method alone does not fully explain the differences found between bracket systems for torque expression and engagement angles. Other aspects of bracket design, such as bracket rigidity and slot depth, in addition to wire manufacturing tolerances, likely contribute to these findings.

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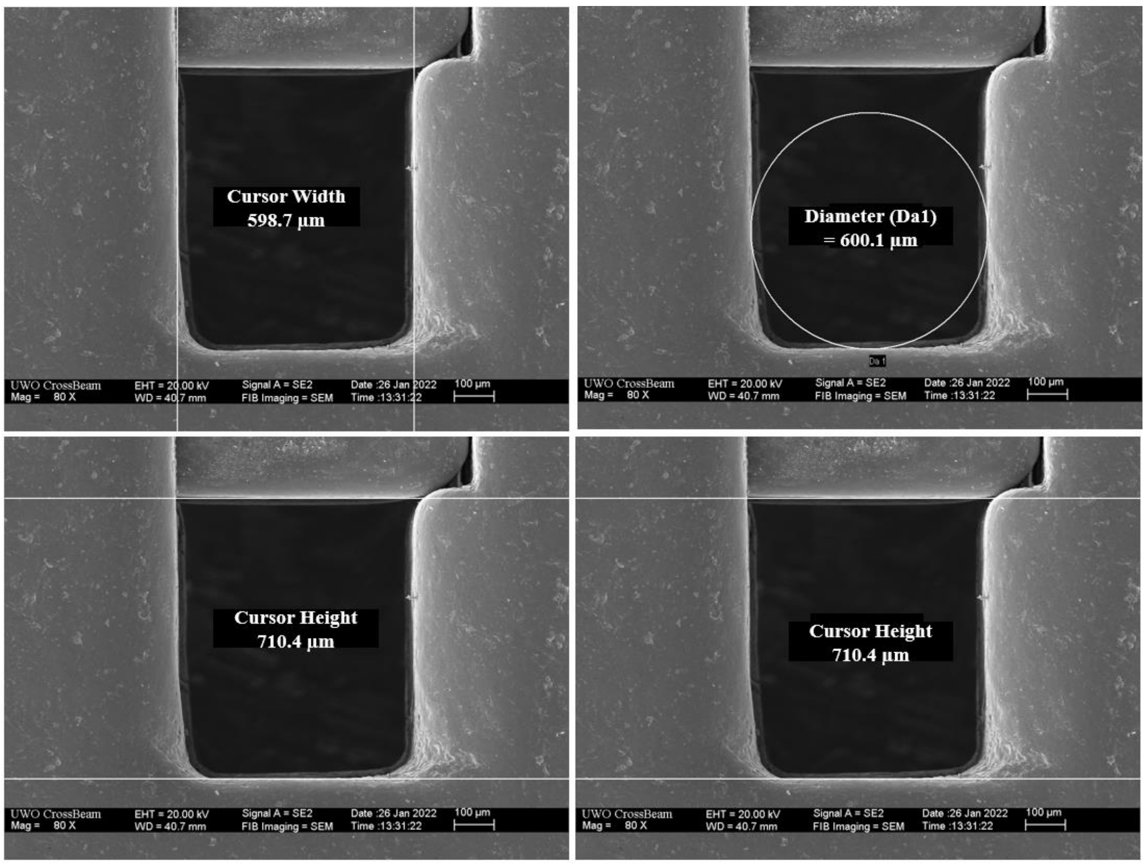
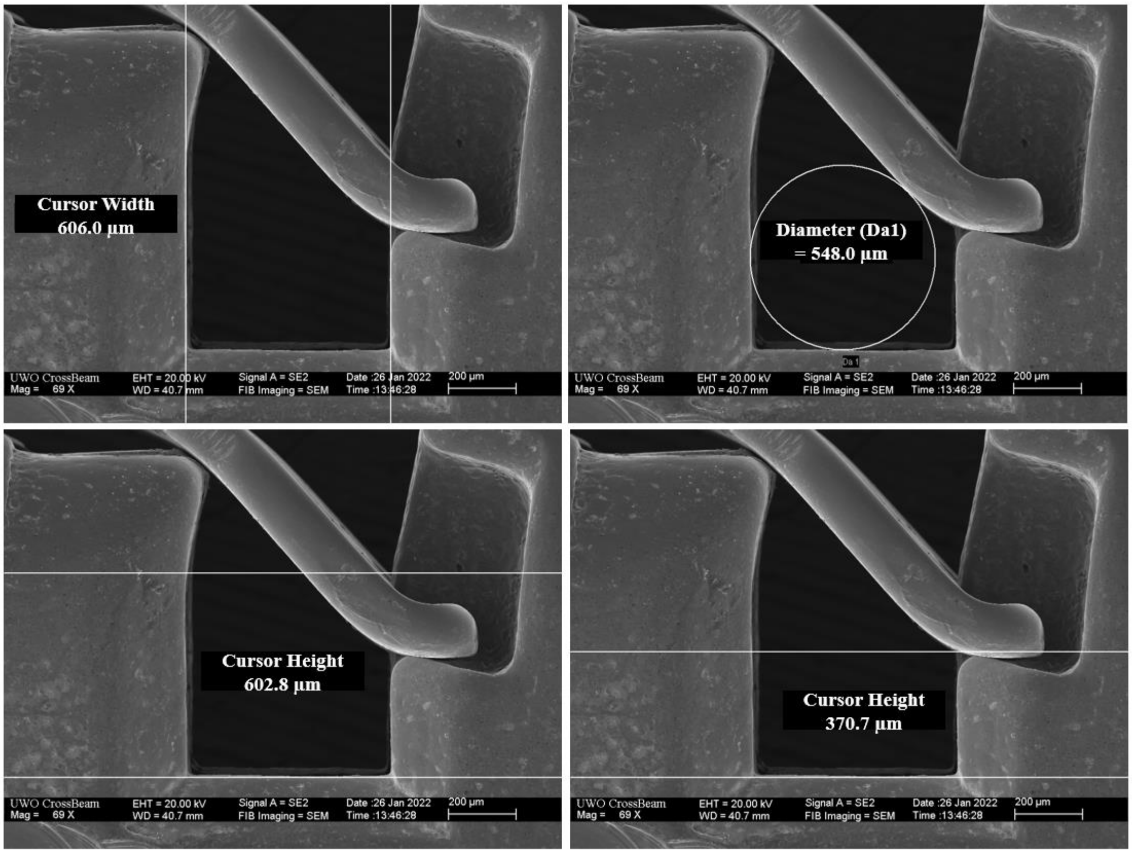
Appendices

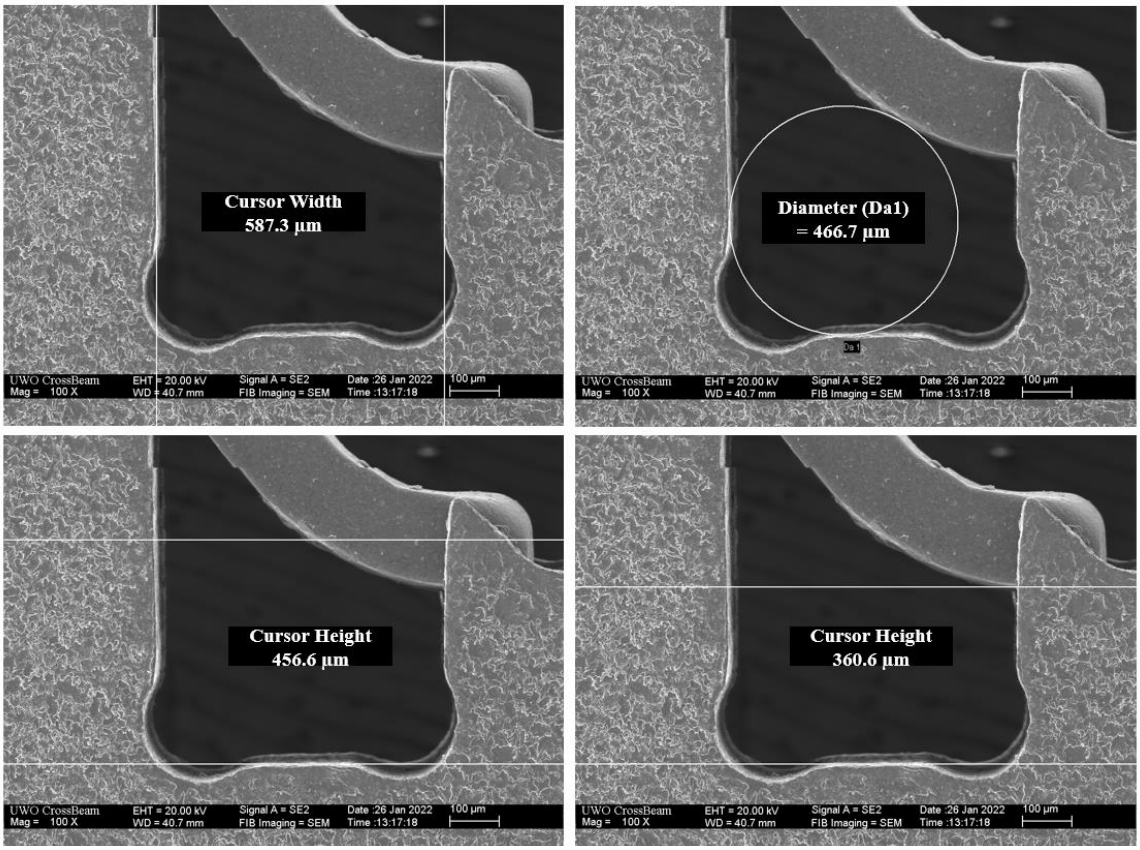
Appendix A: SEM images of P-Dmn brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 100µm = 0.003937-in



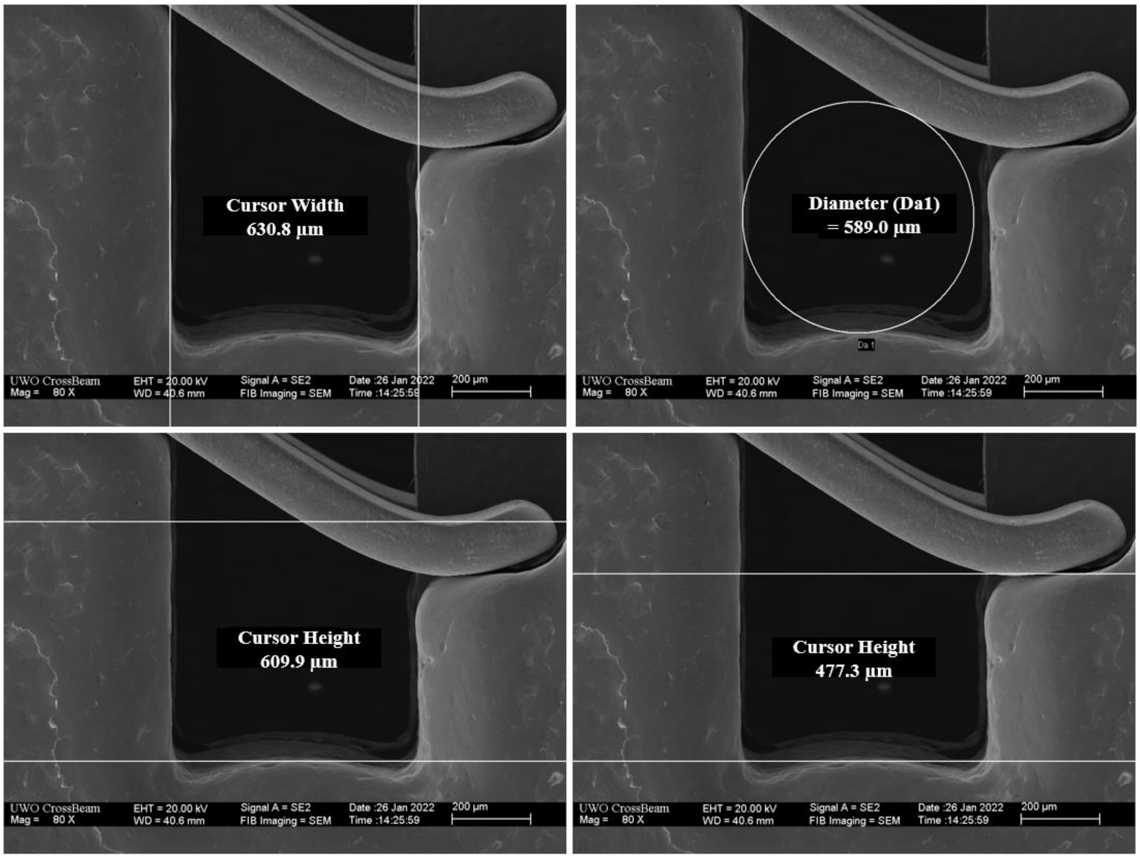
Appendix B: SEM images of A-Emp brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 100µm = 0.003937-in



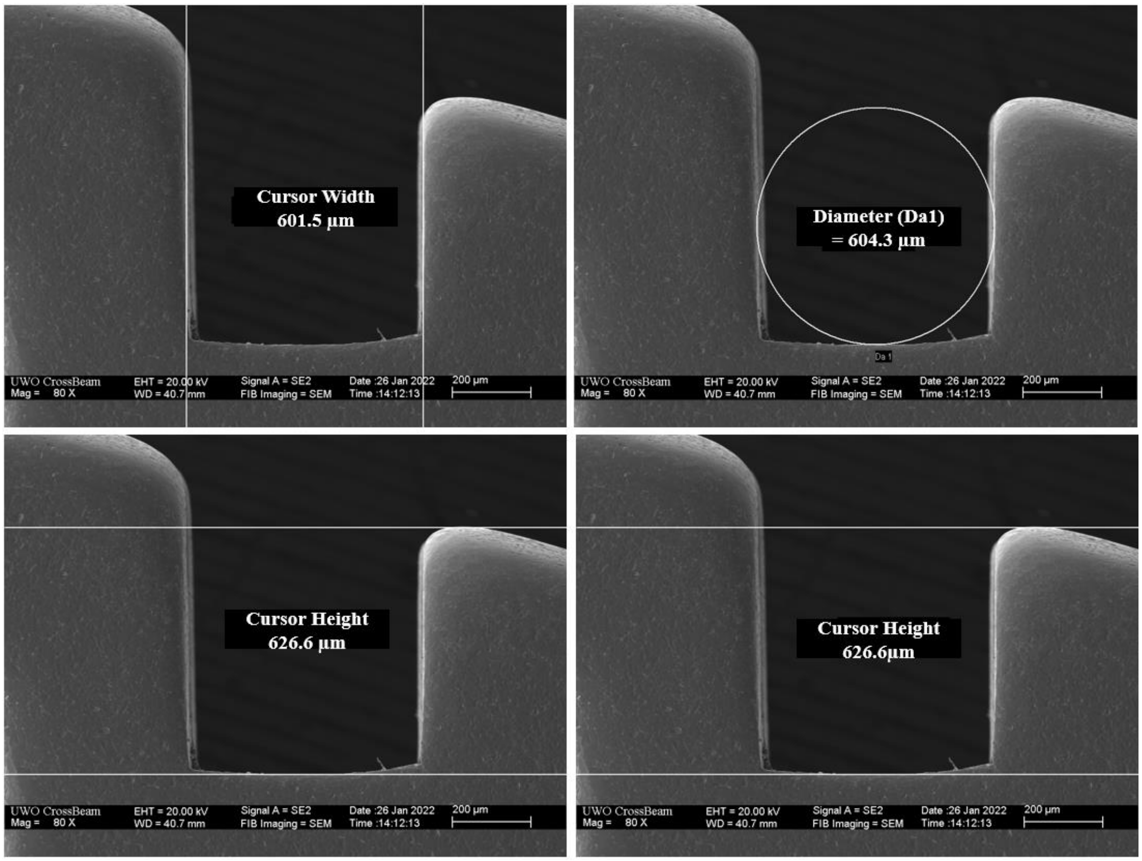
Appendix C: SEM images of A-Spd brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 100µm = 0.003937-in



Appendix D: SEM 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 100µm = 0.003937-in

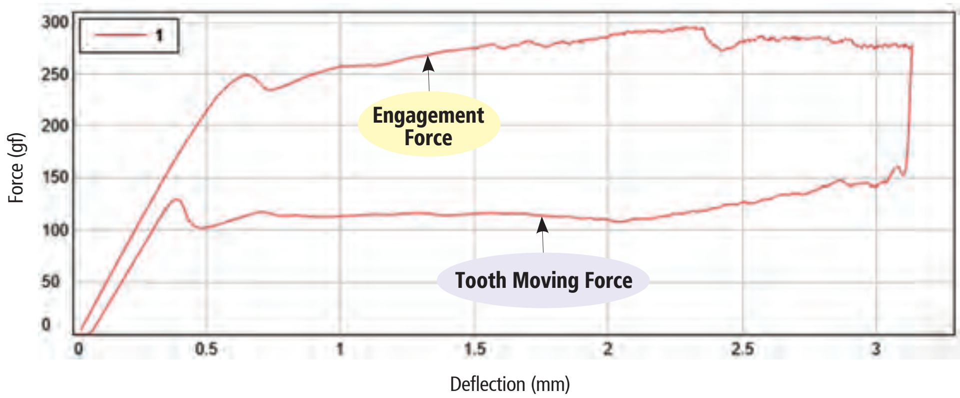


Appendix E: SEM images of C-Vic brackets with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 100µm = 0.003937-in

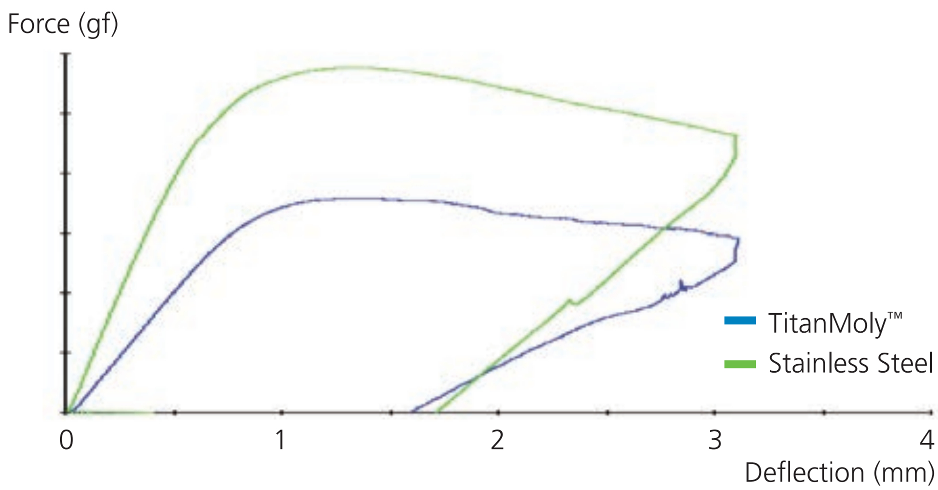


Appendix F: Force-deflection curves of NiTi (A), and TMA and SS (B) wires of interest, as adapted from the manufacturer G&H Orthodontics[70,71]

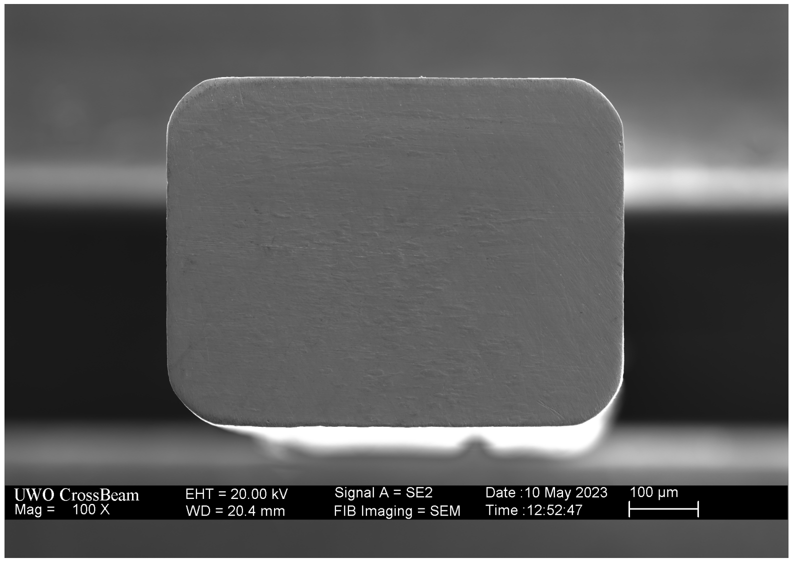
A

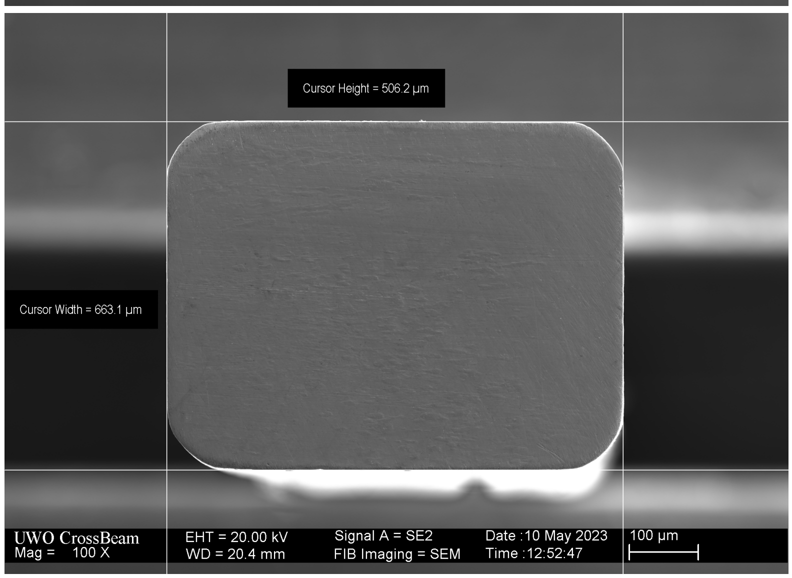


B

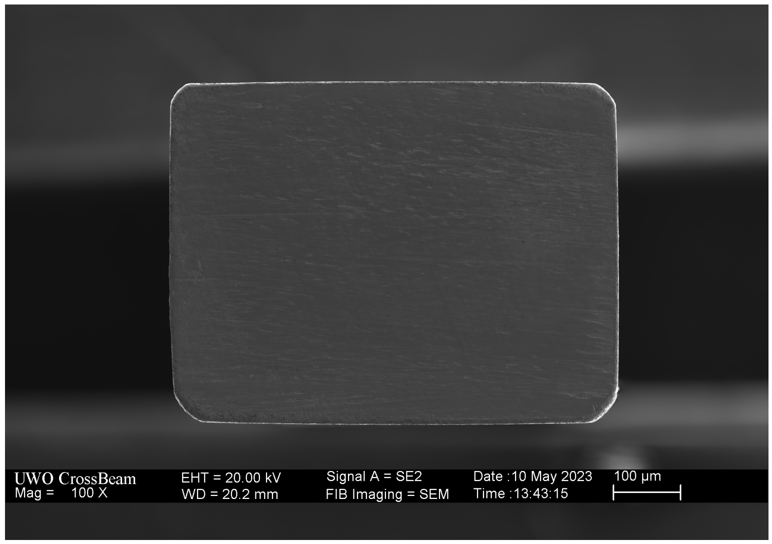


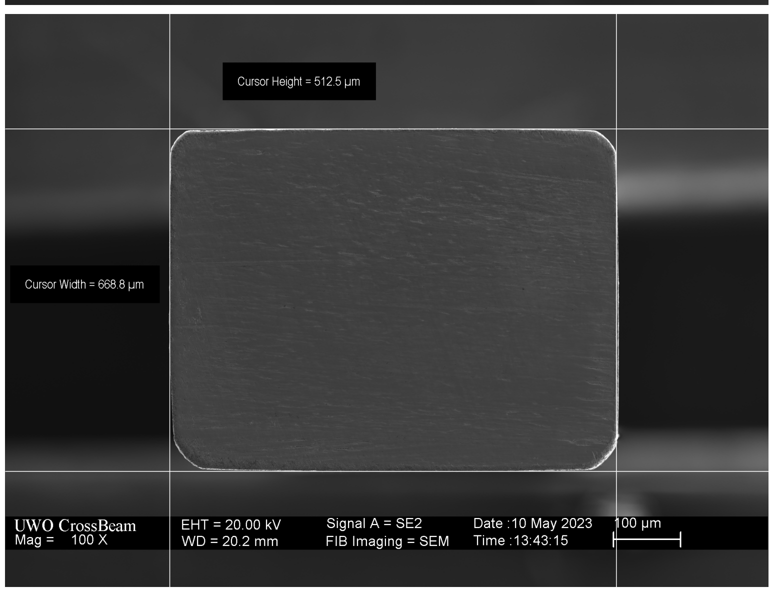
Appendix G: SEM images of NiTi archwire with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 100µm = 0.003937-in. Actual dimensions of wire are 0.0199 x 0.0261-in



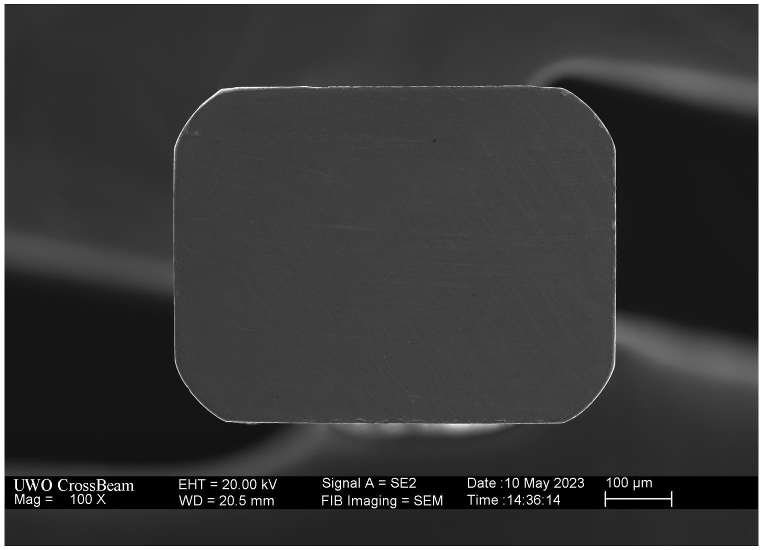


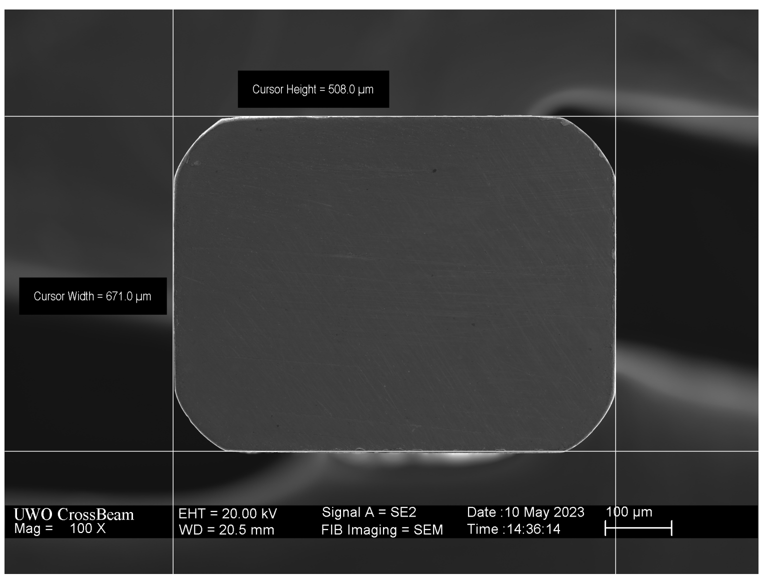
Appendix H: SEM images of TMA archwire with various measurements and included reference points from which measurements were made. Magnification and image scale is included in each individual image. Measures are reported in µm, where 100µm = 0.003937-in. Actual dimensions of wire are 0.0202 x 0.0263-in





Appendix I: SEM images of SS archwire 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 100µm = 0.003937-in. Actual dimensions of wire are 0.0200 x 0.0264-in





Appendix J: Mean torque values (Nmm) for every 3 degrees of rotation for each bracket group tested with NiTi wires

	Bracket System									
Angle (°)	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	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.1	0.0	0.1
3	0.4	0.7	0.3	0.6	0.2	0.4	0.2	0.2	0.3	0.4
6	0.9	1.0	0.4	0.8	0.5	0.5	0.4	0.3	0.6	0.6
9	2.2	1.4	0.8	0.7	1.5	0.8	1.1	0.6	1.4	0.8
12	3.9	2.2	2.1	0.5	2.9	1.0	2.3	0.9	3.1	1.5
15	5.7	3.4	3.8	0.9	4.6	1.2	3.9	1.1	4.9	2.5
18	7.9	5.0	5.9	2.1	6.6	2.1	6.0	1.6	7.3	3.8
21	10.0	6.7	8.2	3.5	8.8	3.1	8.0	2.4	9.6	5.4
24	12.1	8.7	10.3	5.0	11.0	4.3	10.0	3.4	11.8	7.1
27	14.3	10.9	13.0	6.9	13.4	5.8	12.2	4.8	14.3	9.3
30	16.1	13.2	15.4	8.8	15.5	7.3	14.1	6.1	16.3	11.5
33	17.5	15.2	17.1	10.7	17.3	8.9	15.7	7.5	17.8	14.0
36	18.5	16.8	18.6	12.7	18.6	10.6	17.1	9.4	19.1	16.3
39	19.3	18.0	19.6	14.4	19.6	12.2	18.1	11.2	19.8	17.8
42	19.8	18.8	20.2	15.9	20.3	13.8	18.8	13.1	20.3	19.0
45	20.2	19.3	20.8	17.0	20.8	15.1	19.3	15.1	20.7	19.8
42'	17.4	16.7	17.7	14.7	18.1	12.8	16.7	12.7	17.7	17.0
39'	15.2	14.6	15.5	12.4	15.9	10.6	14.5	10.4	15.5	14.6
36'	13.3	12.7	13.5	10.4	14.0	8.5	12.7	8.3	13.7	12.6
33'	11.8	11.2	11.9	8.7	12.5	6.7	11.3	6.5	12.3	11.0
30'	10.7	10.0	10.7	6.9	11.3	5.2	10.2	5.0	11.1	9.5
27'	9.7	8.7	9.5	5.2	10.1	4.0	9.0	3.8	10.1	7.9
24'	8.6	7.4	8.0	3.5	8.8	3.0	7.8	2.6	9.0	6.1
21'	7.5	5.9	6.3	2.2	7.3	2.2	6.3	1.7	7.7	4.6
18'	6.1	4.3	4.3	1.1	5.5	1.4	4.5	1.1	6.1	3.3
15'	4.3	2.8	2.4	-0.1	3.6	0.9	2.7	0.7	4.0	2.2
12'	2.6	1.6	0.9	-0.8	2.1	0.6	1.4	0.5	2.1	1.1
9'	1.0	0.9	-0.4	-0.9	0.8	0.5	0.5	0.3	0.7	0.3
6'	-0.1	0.6	-1.0	-0.8	0.1	0.3	0.1	0.0	-0.1	0.0
3'	-0.3	0.4	-0.9	-0.7	-0.2	0.0	0.0	-0.1	-0.3	-0.2
0'	-0.6	-0.3	-1.0	-0.7	-0.4	-0.2	-0.2	-0.3	-0.5	-0.4

Appendix K: Mean torque values (Nmm) for every 3 degrees of rotation for each bracket group tested with TMA wires

Bracket System										
Angle (°)	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	0.1	-0.5	0.0	0.0	-0.1	0.0	0.0	0.0	0.0	0.0
3	1.1	1.0	0.4	0.4	0.2	0.2	0.7	0.3	0.4	0.3
6	2.6	3.0	0.9	0.2	0.9	0.6	1.6	0.5	0.7	0.7
9	4.9	6.0	3.0	-0.1	2.6	1.4	4.1	0.8	2.1	2.0
12	7.8	9.3	5.5	1.1	4.5	2.9	7.1	1.6	4.1	3.9
15	11.1	12.8	8.2	3.0	7.0	4.8	10.4	3.0	6.9	6.3
18	15.0	17.2	11.8	5.6	10.4	6.9	14.1	5.6	10.7	9.9
21	19.4	21.8	15.8	8.3	13.9	9.1	17.9	8.4	14.7	13.5
24	24.1	26.5	20.1	11.0	17.7	11.2	22.1	11.3	19.1	17.6
27	29.1	31.4	25.2	14.6	22.4	13.6	26.8	14.6	24.5	22.8
30	34.0	36.2	30.0	18.3	26.7	15.9	31.5	17.9	29.5	27.6
33	39.1	41.0	34.5	22.2	31.1	18.3	36.2	21.5	34.5	32.5
36	44.1	45.8	39.6	26.5	36.0	21.4	41.1	25.6	40.0	37.6
39	48.8	50.2	43.8	30.8	40.4	24.2	45.7	29.5	44.8	42.6
42	53.1	54.1	48.4	34.5	44.8	26.7	50.0	33.5	49.2	47.1
45	56.9	57.5	52.7	38.5	49.1	29.5	54.0	37.6	53.7	51.5
42'	52.0	52.5	47.9	34.1	44.5	25.4	49.3	33.2	48.5	46.5
39'	46.4	47.2	42.7	29.1	39.5	20.9	44.0	28.4	43.1	41.0
36'	40.9	41.6	37.2	23.7	34.3	16.6	38.3	23.5	37.4	35.5
33'	35.5	36.4	31.8	19.2	29.3	13.2	33.1	18.9	31.7	29.8
30'	30.3	31.3	26.5	14.9	24.5	10.2	27.9	15.0	26.5	24.5
27'	25.0	26.1	21.7	11.0	20.0	7.5	22.9	11.4	21.4	19.6
24'	20.1	21.1	16.7	7.7	15.5	5.2	18.4	8.0	16.3	14.7
21'	15.6	16.4	12.5	4.9	11.6	3.5	14.3	5.1	12.1	11.0
18'	11.6	12.4	8.7	2.3	8.2	2.3	10.6	3.0	8.4	7.5
15'	8.0	8.8	5.4	0.7	5.0	1.4	7.1	1.6	4.7	4.2
12'	4.7	5.6	2.6	-0.8	2.6	0.9	4.0	1.0	2.0	1.8
9'	2.1	2.6	0.1	-1.3	1.1	0.5	1.6	0.7	0.6	0.4
6'	0.4	0.1	-1.4	-1.2	0.1	0.2	0.4	0.4	-0.1	-0.2
3'	-0.3	-0.9	-1.3	-1.0	-0.3	-0.1	-0.1	0.2	-0.4	-0.5
0'	-0.8	-1.3	-1.4	-1.0	-0.5	-0.3	-0.3	0.0	-0.7	-0.8

Appendix L: Mean torque values (Nmm) for every 3 degrees of rotation for each bracket group tested with SS wires

	Bracket System									
Angle (°)	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	0.0	0.0	0.0	0.0	-1.1	0.0	0.0	0.0	0.0	0.3
3	0.7	0.7	0.4	0.9	-0.7	0.3	0.4	0.1	0.3	0.6
6	3.1	1.8	0.6	1.2	-0.2	0.5	1.0	0.4	0.6	0.9
9	7.6	4.7	1.1	1.3	0.7	0.8	3.4	0.6	1.8	1.2
12	12.8	9.2	3.8	1.4	3.8	1.8	7.2	0.8	5.4	3.0
15	18.0	14.1	8.2	2.8	8.1	3.7	11.9	1.2	9.9	7.8
18	24.1	19.8	13.7	6.6	13.9	6.5	17.3	2.9	15.9	13.4
21	29.9	25.6	19.2	10.7	19.6	9.5	22.5	5.4	22.0	18.7
24	36.6	32.3	24.8	15.0	25.0	12.3	28.1	8.5	28.5	24.2
27	45.0	40.2	31.9	19.5	32.0	15.5	34.7	12.6	36.8	30.3
30	52.2	47.7	39.3	23.9	38.8	18.8	41.6	16.7	44.7	37.2
33	59.9	54.8	46.7	28.4	46.1	22.0	48.6	21.1	52.2	44.5
36	67.5	62.3	54.2	33.2	53.5	25.6	55.5	25.7	60.2	51.9
39	73.5	68.7	60.8	37.9	59.4	29.0	61.7	30.4	67.2	58.7
42	79.5	74.4	66.8	42.4	64.0	31.8	67.6	35.6	73.1	65.1
45	84.8	79.9	72.5	46.6	64.6	34.3	73.0	41.0	79.1	71.1
42'	74.2	69.7	62.2	37.3	53.6	26.5	63.0	32.6	68.1	60.9
39'	64.2	60.0	52.3	28.6	44.7	19.7	53.2	24.8	57.6	50.8
36'	54.3	50.1	41.8	20.9	35.7	14.0	43.4	17.8	47.0	40.6
33'	44.5	40.7	32.4	14.4	27.9	9.6	34.3	12.1	37.2	31.2
30'	35.5	31.9	24.2	8.9	21.3	6.5	26.4	8.1	28.2	23.6
27'	27.0	24.2	17.4	4.8	15.7	4.6	19.8	5.4	20.7	17.2
24'	20.1	17.8	11.5	2.2	10.8	3.1	14.1	3.6	14.3	11.4
21'	14.1	12.4	6.6	0.8	6.7	2.1	9.3	2.1	9.4	7.0
18'	8.4	7.3	2.7	0.2	3.2	1.5	5.6	1.3	5.3	3.7
15'	3.6	3.4	0.4	0.1	1.0	1.0	2.6	0.8	2.0	1.2
12'	0.9	1.0	-0.9	0.0	-0.3	0.6	0.8	0.6	0.3	0.5
9'	-0.3	0.4	-1.2	0.0	-0.6	0.4	0.2	0.4	-0.2	0.2
6'	-0.6	0.1	-1.2	-0.1	-0.8	0.2	-0.2	0.2	-0.3	0.0
3'	-0.9	-0.3	-1.2	-0.3	-1.1	0.0	-0.4	0.0	-0.5	-0.3
0'	-1.4	-1.2	-1.4	-0.9	-1.6	-0.3	-0.6	-0.2	-0.8	-0.7

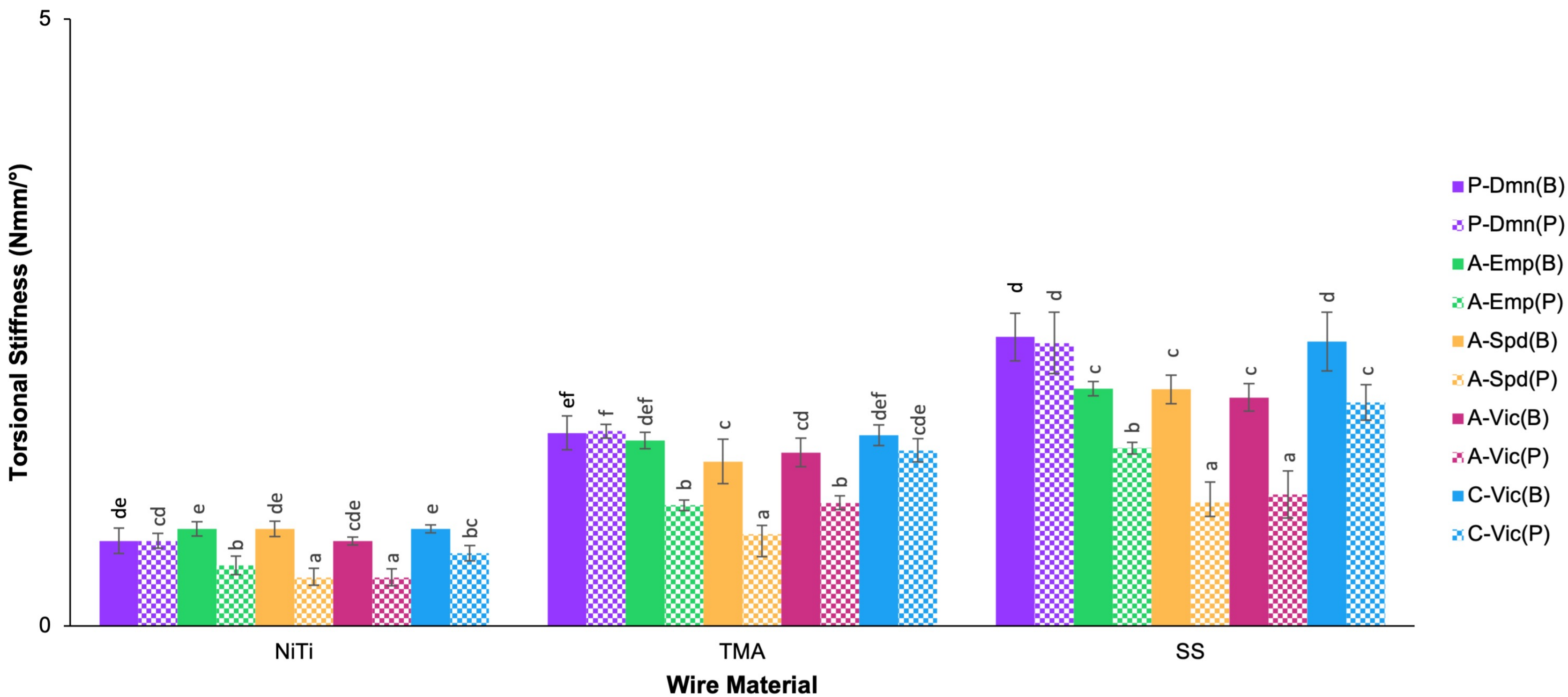
Appendix M: Mean torsional stiffness in Nmm/° (±SD) for each bracket system-wire combination, as measured on the loading curve between 20 and 25 degrees of rotation. Non-significant differences between bracket systems within a wire material at P>0.05 are denoted by shared alphabetical letters within each row

	Bracket Group									
Wire Material	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)
NiTi	0.7(0.1)^{de}	0.7(0.1)^{cd}	0.8(0.1)^{e}	0.5(0.1)^{b}	0.8(0.1)^{de}	0.4(0.1)^{a}	0.7(0.03)^{cde}	0.4(0.1)^{a}	0.8(0.03)^{e}	0.6(0.1)^{bc}
TMA	1.6(0.1)^{ef}	1.6(0.1)^{f}	1.5(0.1)^{def}	1.0(0.04)^{b}	1.4(0.2)^{c}	0.8(0.1)^{a}	1.4(0.1)^{cd}	1.0(0.1)^{b}	1.6(0.1)^{def}	1.4(0.1)^{cde}
SS	2.4(0.2)^{d}	2.3(0.1)^{d}	2.0(0.1)^{c}	1.5(0.1)^{b}	1.9(0.1)^{c}	1.0(0.2)^{a}	1.9(0.01)^{c}	1.1(0.2)^{a}	2.3(0.2)^{d}	1.8(0.5)^{c}

Significance results comparing mean torsional stiffness between bracket systems within a wire material

Wire Material	Significance Results
NiTi	$F(9,90) = 50.05$, $P<0.001$
TMA	$F(9,90) = 87.82$, $P<0.001$
SS	$F(9,90) = 87.42$, $P<0.001$

Appendix N: Mean torsional stiffness in Nmm/° (±SD) with different bracket systems versus wire material, as measured on the loading curve between 20 and 25 degrees of rotation. Error bars represent 1 SD, and letters shared within each wire material cluster represent non-significant differences between bracket systems at P>0.05



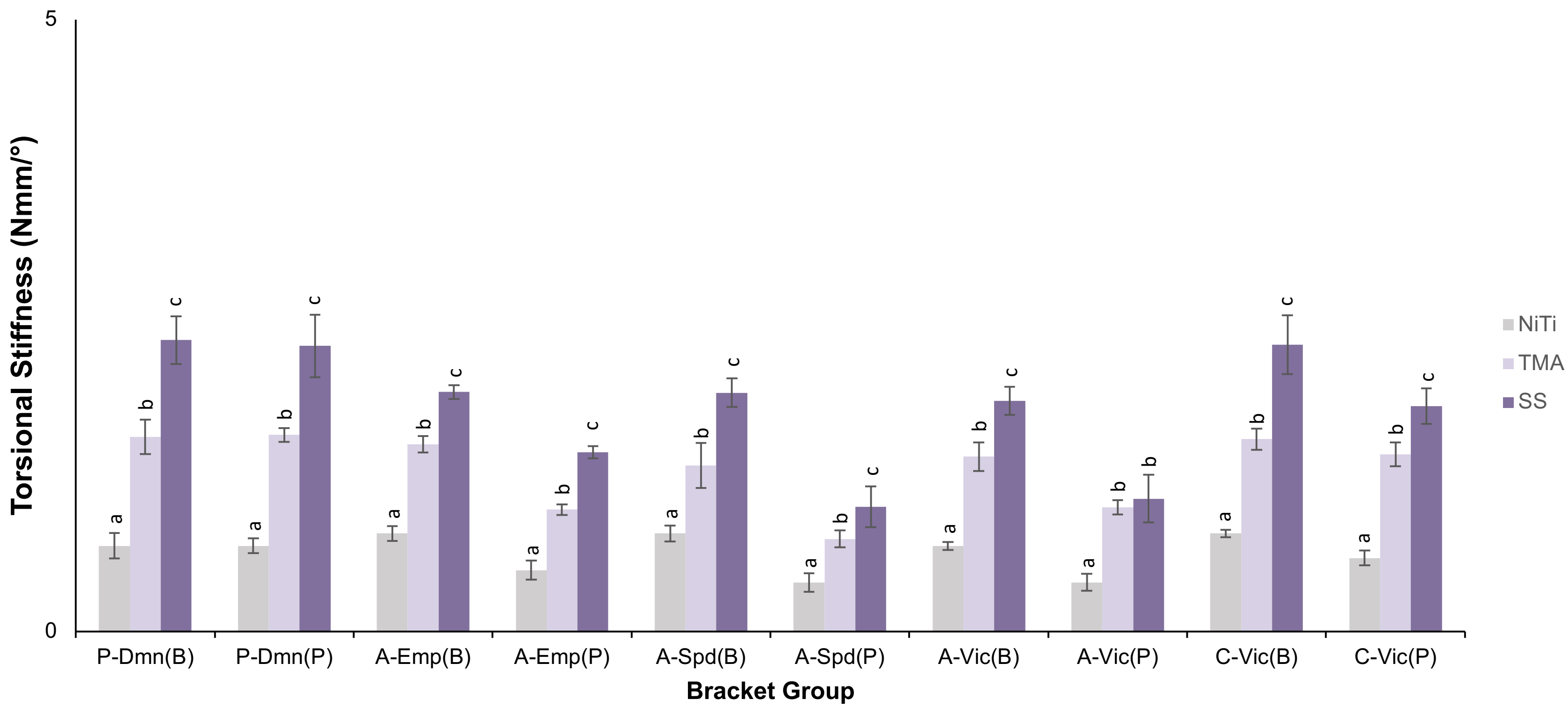
Appendix O: Mean torsional stiffness in Nmm/° (±SD) for each bracket system-wire combination, as measured on the loading curve between 20 and 25 degrees of rotation. Non-significant differences between wire materials within a given bracket system at P>0.05 are denoted by shared alphabetical letters within each row

	Wire Material		
Bracket Group	NiTi	TMA	SS
P-Dmn(B)	0.7(0.1)[a]	1.6(0.1)[b]	2.4(0.2)[c]
P-Dmn(P)	0.7(0.1)[a]	1.6(0.1)[b]	2.3(0.3)[c]
A-Emp(B)	0.8(0.1)[a]	1.5(0.1)[b]	2.0(0.1)[c]
A-Emp(P)	0.5(0.1)[a]	1.0(0.04)[b]	1.5(0.05)[c]
A-Spd(B)	0.8(0.1)[a]	1.4(0.2)[b]	1.9(0.1)[c]
A-Spd(P)	0.4(0.1)[a]	0.8(0.1)[b]	1.0(0.2)[c]
A-Vic(B)	0.7(0.03)[a]	1.4(0.1)[b]	1.9(0.1)[c]
A-Vic(P)	0.4(0.1)[a]	1.0(0.1)[b]	1.1(0.2)[b]
C-Vic(B)	0.8(0.03)[a]	1.6(0.1)[b]	2.3(0.2)[c]
C-Vic(P)	0.6(0.1)[a]	1.4(0.1)[b]	1.8(0.1)[c]

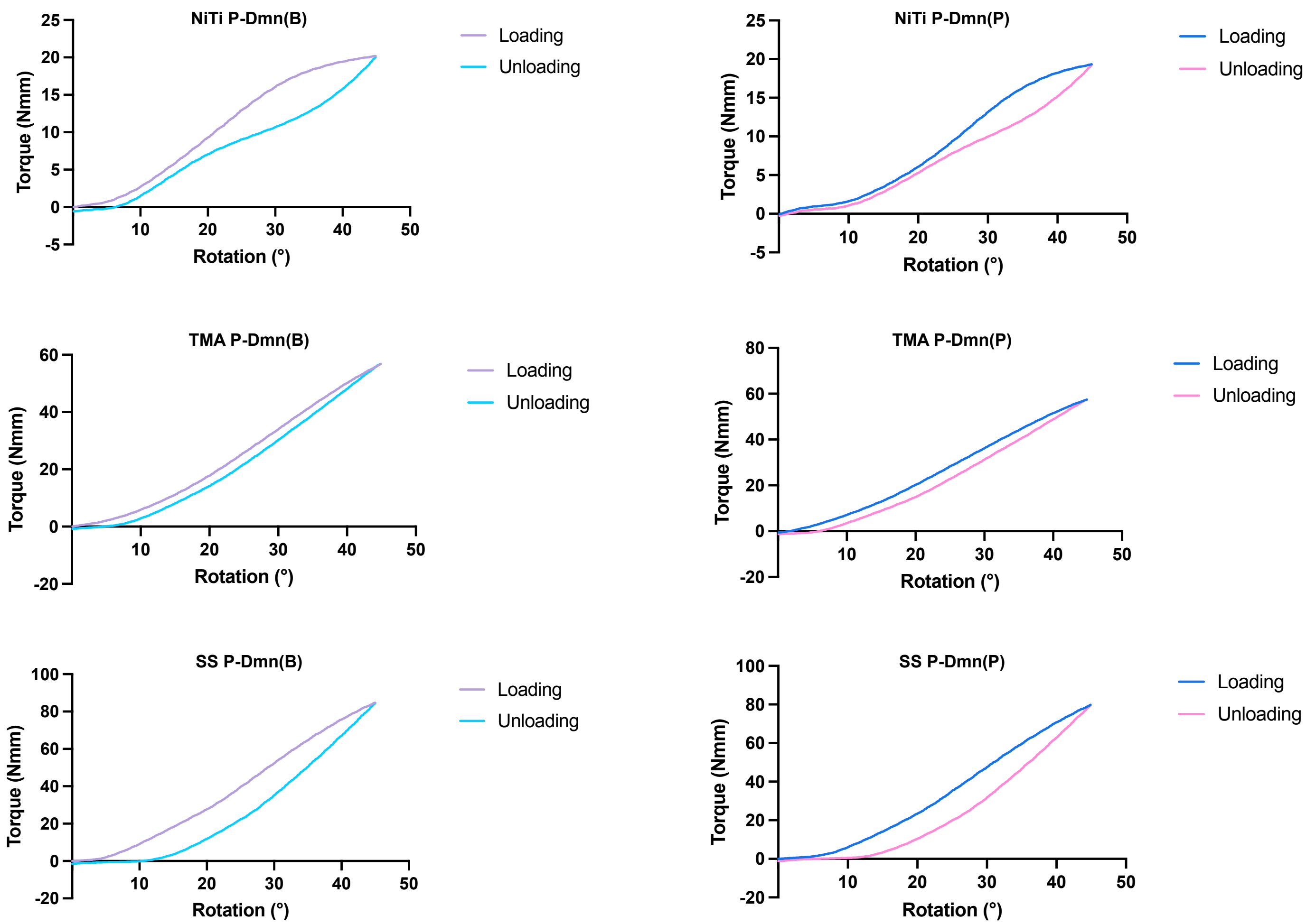
Significance results comparing mean torsional stiffness between wire materials within a given bracket group

Bracket Group	Significance Results
P-Dmn(B)	$F(2,27) = 298.98, P<0.001$
P-Dmn(P)	$F(2,27) = 291.25, P<0.001$
A-Emp(B)	$F(2,27) = 915.41, P<0.001$
A-Emp(P)	$F(2,27) = 626.76, P<0.001$
A-Spd(B)	$F(2,27) = 205.08, P<0.001$
A-Spd(P)	$F(2,27) = 66.85, P<0.001$
A-Vic(B)	$F(2,27) = 380.35, P<0.001$
A-Vic(P)	$F(2,27) = 100.29, P<0.001$
C-Vic(B)	$F(2,27) = 270.62, P<0.001$
C-Vic(P)	$F(2,27) = 345.71, P<0.001$

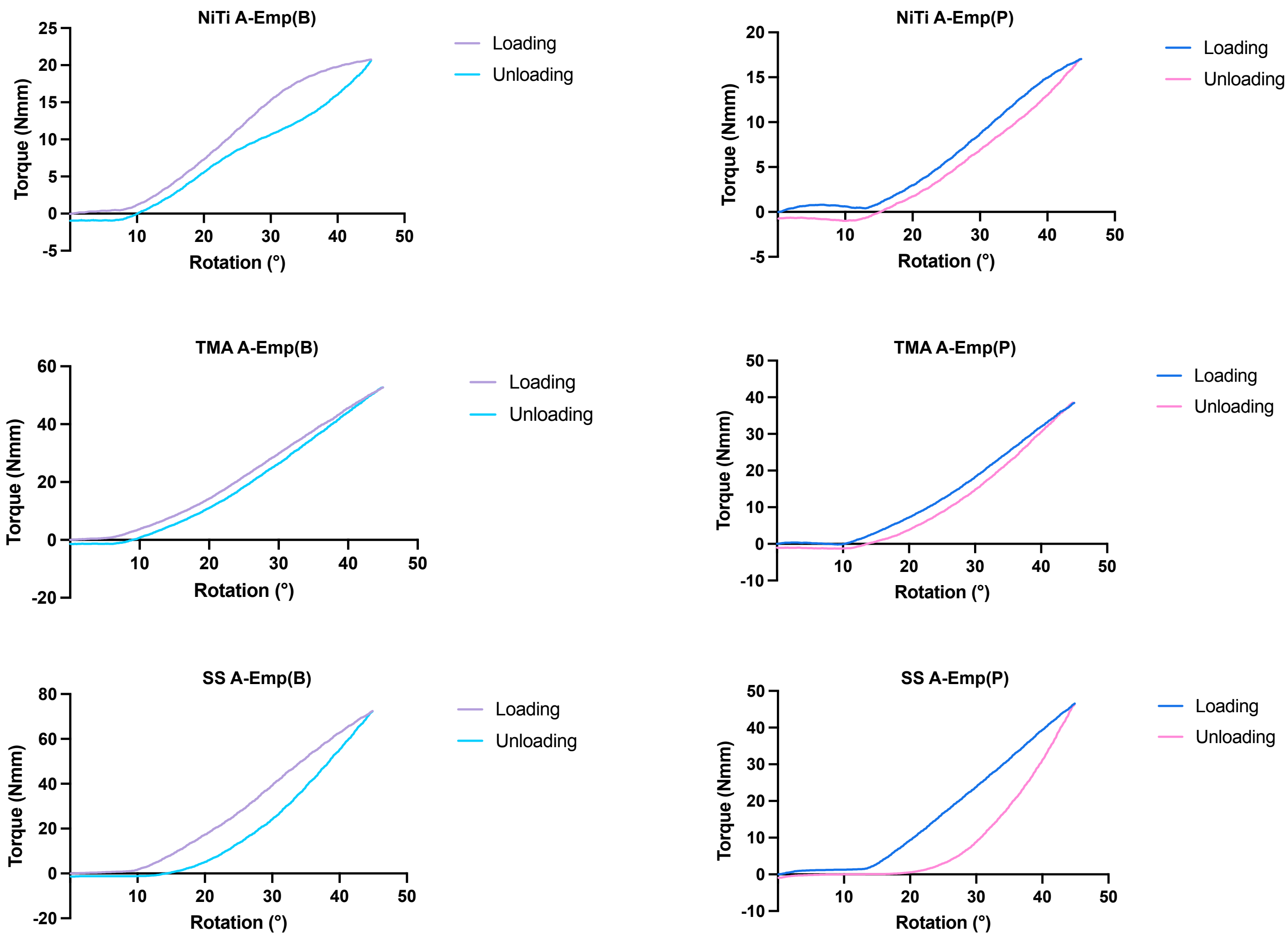
Appendix P: Mean torsional stiffness in Nmm/° (±SD) with different wire materials versus bracket systems, as measured on the loading curve between 20 and 25 degrees of rotation. Error bars represent 1 SD, and letters shared within each bracket system cluster represent non-significant differences between wire materials at P>0.05



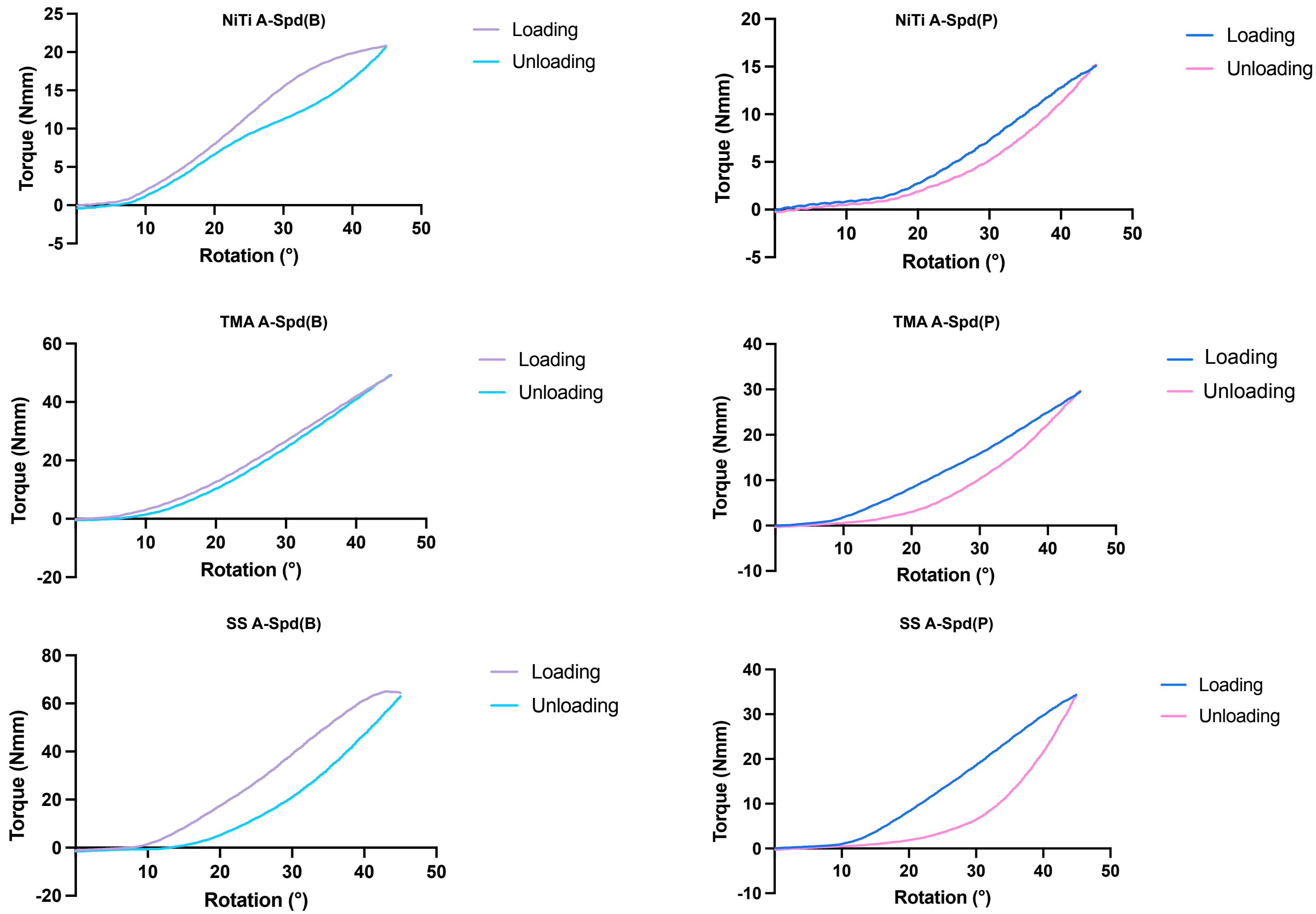
Appendix Q: Mean torque-rotation loading and unloading curves of P-Dmn, for all wire materials and directions of rotation



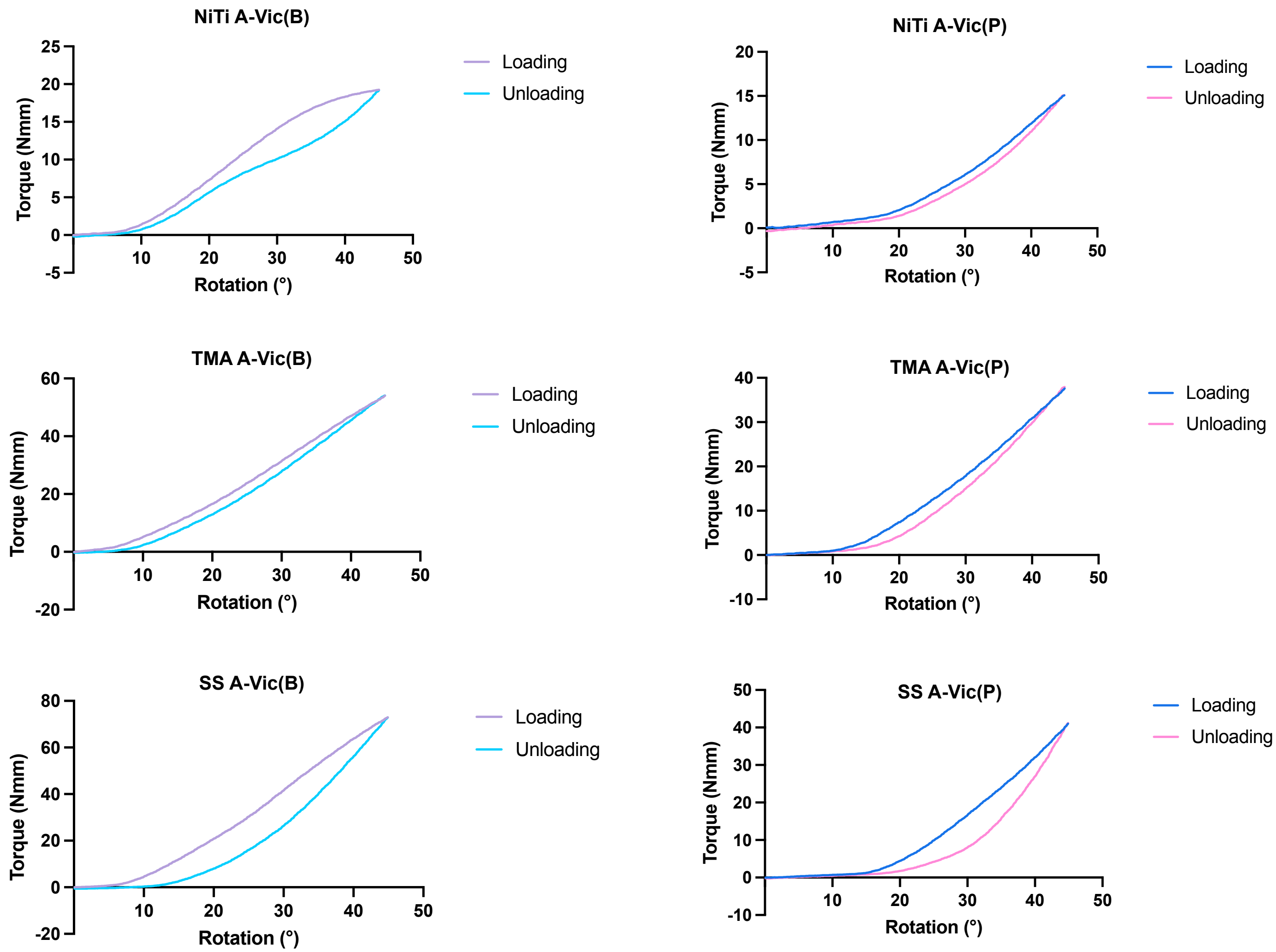
Appendix R: Mean torque-rotation loading and unloading curves of A-Emp, for all wire materials and directions of rotation



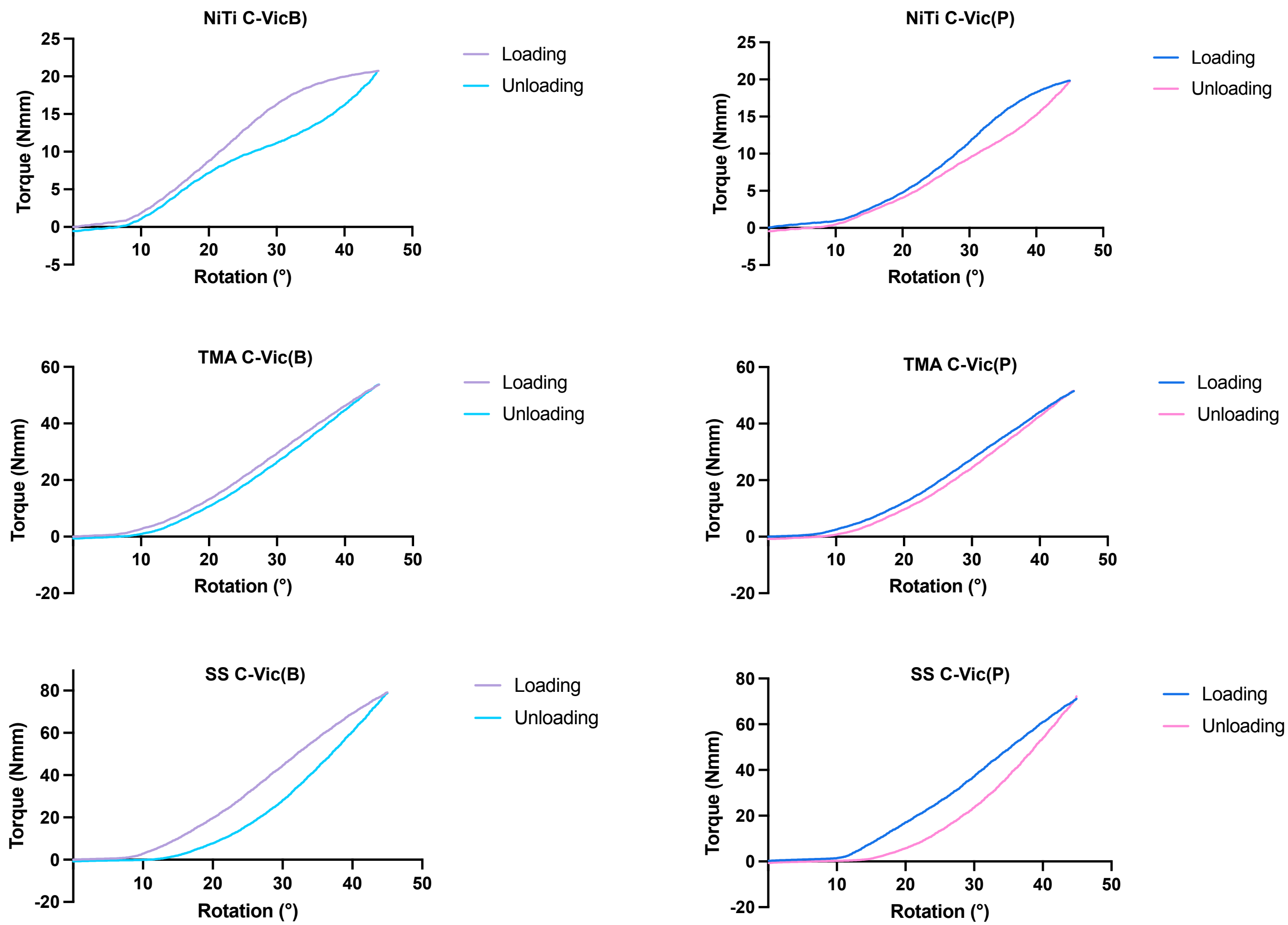
Appendix S: Mean torque-rotation loading and unloading curves of A-Spd, for all wire materials and directions of rotation



Appendix T: Mean torque-rotation loading and unloading curves of A-Vic, for all wire materials and directions of rotation



Appendix U: Mean torque-rotation loading and unloading curves of C-Vic, for all wire materials and directions of rotation



Curriculum Vitae

Name:	Sidney Mugford
Post-secondary Education and Degrees:	University of Western Ontario London, Ontario, Canada 2020-2023 MClD (Orthodontics)
	Dalhousie University Halifax, Nova Scotia, Canada 2013-2017 DDS
	Cape Breton University Sydney, Nova Scotia, Canada 2009-2013 BSc (Biology)
Honours and Awards:	Dr. David C. Way Memorial Award 2021
	International Collage of Dentists Award 2016
Related Work Experience	Teaching Assistant The University of Western Ontario 2022
	Associate Dentist David Quigley Dentistry 2019-2020
	Associate Dentist Bridgeview Dental 2018-2019
	Associate Dentist Westmount Dental Clinic 2017-2018