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Torque Expression of Contemporary Self-ligating Bracket Systems

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Abstract

Background: Self-ligating bracket systems in orthodontics have evolved over the years, but there is limited data regarding their ability to generate torque for efficient third-order tooth positioning.

Aim: To compare contemporary active self-ligating bracket systems against a passive self-ligating bracket system and a traditional twin bracket system in their ability to generate torque at different degrees and direction of wire rotation, in vitro.

Materials and Methods: Five bracket system groups of 0.022 inch slot size (twin bracket system Victory Series with elastic ligature [E-Vic]; passive self-ligating bracket system Damon Q [P-Dmn]; and active self-ligating bracket systems Speed [A-Spd], InOvation-R [A-Ovn] and Empower 2 Active[A-Emp]) were tested for torque expression utilizing a 0.019 x 0.025 inch stainless steel wire ligated into their slots. Single upper right central incisor brackets of each system were mounted using a specialized mounting jig, and a custom torque assembly fixed to an Instron materials testing machine was utilized to measure torque generated from -15 to +45 degrees of wire rotation. Ten clockwise and ten counterclockwise rotations were performed for each bracket system (n=20).

Results: Torque expression significantly varied between bracket systems with P-Dmn, E-Vic and A-Ovn generating the highest torque, and A-Spd and A-Emp the lowest, at most degrees of wire rotation. The direction of wire rotation had the largest effect on the A-Spd and A-Emp active bracket systems, whereby the counterclockwise rotation generated significantly more torque than the clockwise rotation tests.

Conclusions: All five bracket systems displayed different behaviors of torque expression when comparing degrees and direction of wire rotation. Understanding these differences in torque expression can help the clinician plan and provide treatment more efficiently.

Keywords

Orthodontic Brackets, Torque, Torque Expression, Torque Play, Passive Self-Ligation, Active Self-Ligation

Summary for Lay Audience

For many reasons, it is important to properly position teeth within the mouth. Many different bracket systems are commonly used in orthodontics to move the front teeth as desired. These different systems often have varying ways to secure the archwire to the bracket, resulting in different forces generated. This study was designed to test the ability of five different contemporary orthodontic bracket systems to produce torque. A custom set-up was made to twist a commonly used archwire inside of an orthodontic bracket. This was done from -15 to +45 degrees for each of the five bracket systems, in both the clockwise and counterclockwise directions. Results were compared to one another and other studies to evaluate how efficient each group was at producing torque in both directions.

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List of Abbreviations

ANOVA	Analysis of Variance
A-Emp	Empower 2 Active
A-Ovn	InOvation-R
A-Spd	Speed
CAD/CAM	Computer Aided Design and Manufacturing
E-Vic	Victory Series
ES	Effect Size
-in	Inch
Nmm	Newton Millimeters
NT	Nickel Titanium
OMSS	Orthodontic Measuring and Simulation System
P-Dmn	Damon Q
SEM	Scanning Electron Microscopy
SLB	Self-ligating Bracket
SS	Stainless Steel
SD	Standard Deviation
TMA	Titanium Molybdenum Alloy

Chapter 1

1 Review of the Literature

1.1 Introduction

The orthodontic movements used to improve a patients' smile and occlusion has long been a precise and exacting art within orthodontic treatment. The labiolingual inclination of teeth, also referred to as third-order movements or torque in orthodontics, is an important factor for the esthetics of smiling and also plays an important role in ideal occlusal schemes.[1–3] In order to achieve esthetic and functional goals, orthodontists must have precise control over the third-order movements of the dentition. With this in mind, there are many methods of controlling the movement of teeth and this must be planned before treatment even begins. Choosing particular bracket systems and sizes along with specific prescriptions for the internal slots of these brackets is done before the patient has started active treatment. This is theoretically based on many factors related to the bracket such as the precision of the internal slot component, the material composition and ligation ability.[1, 4–7]

Once treatment has begun, many choices are made with consideration to the wires used. In addition to choosing the most efficient bracket system, the clinician must decide on the appropriate wire size to engage the internal dimensions of these brackets in order to express specific movements of the teeth. Further, deciding on the composition of these wires can influence the capabilities of the bracket-archwire system to produce torque.[8, 9] As readily evident, there are many factors that play into the expression of third-order movements of the incisors. It is therefore of great significance to investigate all of the potential methods used with contemporary bracket systems to affect the torque of these anterior teeth.

Torque is often referred to as a measure of force causing an object to rotate about an axis within the object. It is a complex three-dimensional movement resulting from the forces of a couple around a fulcrum point and can be thought of as the moment that moves an object (Figure 1). In orthodontics, it is often thought of as root movement, as opposed to crown movement which is typically defined by tipping. In the literature, it is typically described in units of Newton millimeters (Nmm) as a direct result of the force magnitudes and minute measurements involved with objects as small as teeth.[1, 2, 7, 10–12]

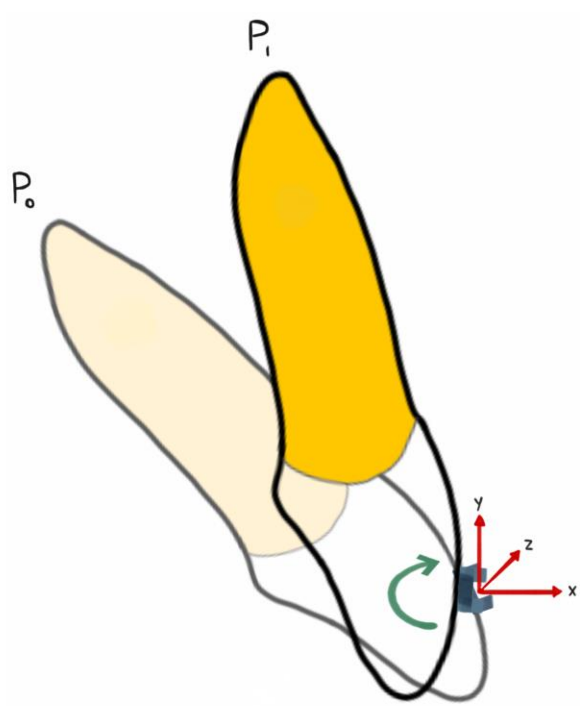


Figure 1: Visual representation of torque in orthodontics.

1.2 Stages of Orthodontic Treatment

The typical sequence of orthodontic treatment when using contemporary straight-wire mechanics involves three separate and distinct phases.[2, 3] Historically, these have been regarded as the levelling and aligning stage, anteroposterior correction stage and finishing stage. The first stage of treatment often involves the use of light, flexible wires for aligning the teeth and levelling them all to be on the same plane of occlusion. At this point in treatment, there is a moderate focus on root movement and the torquing of teeth. The goal of this first phase is to prepare the upper and lower arches to receive larger, straight archwires made of stiffer materials such as stainless steel (SS) or a titanium molybdenum alloy (TMA). As the archwires progress from small round wires to large rectangular wires, the slot slowly becomes actively engaged in the preadjusted appliance. The level of wire engagement in the slot will also vary depending on the method of ligation. The amount of torque built-in to the slot of the bracket is usually fully expressed by the time large rectangular wires of SS or TMA have been in place for a period of time.

Once the teeth have been levelled and aligned with larger wires engaged, the second phase of treatment begins. This phase involves the use of elastics and often other auxiliaries to address discrepancies in the anteroposterior relation of the upper and lower teeth. It also addresses the closure of any remaining spaces that may have resulted from extraction of teeth, functional appliances, or simply from the progression of archwires in the first stage. This second stage often has significant side-effects on the torques of teeth through anteroposterior movements and space closure, such as retroclination of the anterior teeth. Occasionally, specific moments are used in this stage to aid in the appropriate management of space closure through anchorage reinforcement techniques.

After the anteroposterior corrections have been made and all residual spaces are closed, the third stage of finishing begins. At this point in treatment, a very large focus is placed on adjusting any improper third-order positions of the teeth. Along with completing any residual first and second-order movements, clinicians must apply custom torque, beyond that expressed through the bracket prescription, to anterior and posterior teeth as needed. While focusing on esthetics and occlusal interlocking, large rectangular wires are often used to engage the bracket slots intimately and express the amount of torque best suitable for the patient's dentition. Occasionally, auxiliary appliances or devices are needed in order to assist the brackets and wires in achieving ideal torque values for all of the teeth.

1.3 Bracket Systems

Many different bracket systems have been developed since the conception of the orthodontic profession. Each system can differ with regards to many factors, including design (size and shape), materials, built-in prescription, as well as slot size and accuracy, and the form of ligation (Figure 2). Theoretically, the size of the slot and method of ligation will play critical roles in how much torque is expressed between systems with the same size of wire.

Figure 2: Different bracket systems used in this study and commonly used in North America.

1.3.1 Bracket Slots

The size of the bracket slot has a dominant role in how early and readily wires begin to engage the walls of the slot and start producing torque. In orthodontics, these dimensions are typically reported in thousandths of an inch and the two common bracket slot sizes utilized today (height x depth) are 0.018 x 0.025-in (0.46 x 0.64 mm) and 0.022 x 0.028-in (0.56 x 0.71 mm), often referred to as the 0.018-in and 0.022-in systems, respectively.

The advantage of 0.022-in bracket systems comes from the ability to use heavier archwires for stabilization and levelling purposes during treatment.[3, 42] Advocates of the 0.018-in system argue that the smaller internal dimensions allow for more intimate archwire engagement of slot walls, earlier torque expression and greater torque expression with the same sized wires.[5, 48] A recently done systematic review on this subject revealed that there were no significant differences in clinical wire engagement angles between the 0.018-in and 0.022-in systems when testing their largest respective wire sizes (0.017 x 0.025-in wire for the 0.018-in system; 0.021 x 0.025-in for the 0.022-in system).[14]

It must be taken into consideration that the actual versus nominal values for bracket dimensions vary significantly, with the bracket slots often being larger than described by the manufacturer.[1, 4–7, 24, 26, 28, 30, 31, 34] The larger slot size reduces the amount of archwire engagement, although the difference may be as low as 2 degrees from ideal in some cases and therefore not always clinically significant.[7] Size discrepancies may be a result of the manufacturing process which varies between different bracket systems. Metal-injection molding techniques involve the expansion and shrinkage of the bracket materials. Machine milling techniques eliminate this problem of changing dimensions but can leave the brackets with a rough, grainy surface. It has been shown that milled brackets often have metal particles, grooves or striations in the slot walls which effectively change the dimensional accuracy of the slot and may prevent the full engagement of an archwire.[31]

1.3.2 Bracket Engagement Angles

The amount of rotation an archwire must undergo before engaging all three internal walls of the bracket slot is known as the engagement angle, torque play, or bracket slop. The "zero" angle of torque is that in which the wire is centered perfectly in the slot of the bracket and must undergo equal amounts of twist in the positive or negative directions to engage the walls of the bracket slot. As calculated in previous studies using trigonometric calculations with popular working archwires, the theoretical torque play in an 0.022-in bracket with a 0.019 x 0.025-in (0.48 x 0.64 mm) archwire would be 7.2 degrees (Figure 3).[5, 7, 19, 32, 50, 51] Following similar calculations for an 0.018-in bracket with an equivalent 0.016 x 0.022-in (0.41 x 0.56 mm) archwire, the theoretical torque play would be 5.4 degrees.[50] The difference in theoretical slop values between these two systems is readily evident from these calculations.

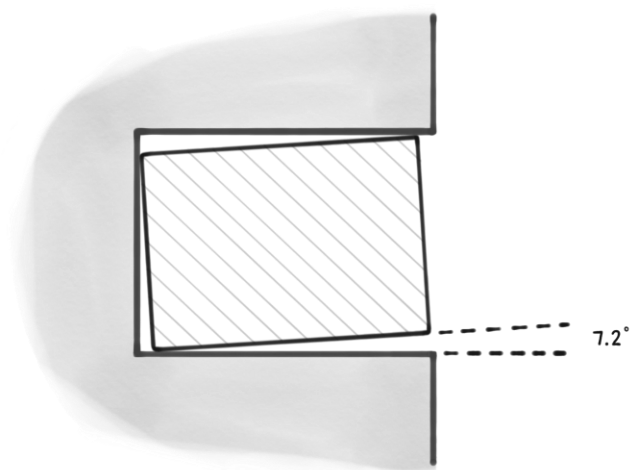


Figure 3: Approximate visualization of the theoretical torque play for a 0.019 x 0.025-in wire in a 0.022 x 0.028-in bracket slot.

The amount of slop on either side of the wire is directly related to the actual dimensions of the brackets and wires, which have both been shown to be quite variable from their manufacturer's nominal values. This leads to the realization that the clinically observed slop in a particular bracket-archwire complex is always different from the theoretical slop calculated from geometry.[1, 5–7, 13, 14, 17, 19, 20, 42, 48, 49] It is important to keep this in mind as the actual slop values always prove to be much greater in literature regarding torquing experiments, due to the many aforementioned factors.

1.3.3 Bracket Prescriptions and Torque

In order to address the differences between actual and theoretical torque, bracket slot variation, and archwire size variation and bevelling, bracket manufacturers fabricate their products with different torque prescriptions. These prescriptions are traditionally based on the ideas of prominent orthodontists worldwide who use different treatment mechanics to move the dentition in unique ways. The maxillary central incisor torque prescription varies from 12 degrees in the Roth prescription, up to 22 degrees in the Ricketts bioprogressive prescription.[2] Some of the more commonly observed prescriptions include that of Andrews, Roth, and MBT.[2] With sound reasoning, these orthodontists developed their bracket prescriptions with regards to issues such as the sagittal relationships of anterior and posterior teeth, total arch length and esthetics related to the inclination of maxillary incisors.

It is believed by many clinicians that brackets with higher built-in torque prescription values produce higher moments on a tooth at any given time, when compared to the same bracket with a lower torque prescription.[5] However, conflicting evidence supports the idea that the slop in any bracket-archwire complex is large enough to negate the torquing prescription differences between different bracket systems.[7] Whether the bracket prescription actually plays a significant role or not, it seems wise to choose one based on what the ideal results would yield with that particular torque value built into the appliance.

1.4 Ligation Methods

The method of ligating the wire to the bracket may differ between systems as well and may involve elastomeric modules, steel ligature ties or sliding doors and clips that are contained within the architecture of the bracket itself (Figure 4 A – C). Manufacturers often propose that their specific method of ligation has an advantage on the ability to move teeth or reduce treatment time.

(A) (B) (C) (D)

Figure 4: Examples of different ligation types: **A)** Elastomeric ligation; **B)** Stainless steel ligature tie ligation; **C)** Passive self-ligation with a closed door; **D)** Passive self-ligation with an open door.

1.4.1 Elastomeric Ligation

Fixed appliances in orthodontics were originally developed to support the use of steel ligation methods. Traditionally, stainless steel (SS) ties were used as the primary method of ligation, but due to the desire for quicker and more user-friendly methods, elastomeric ligation was realized. Elastomers are circular modules made out of elastomeric material through injection molding or cutting techniques. They were developed to press the wire against the base of the bracket and stay attached due to the design of the bracket tie-wings. These elastomeric modules have proved more efficient than traditional SS ties and are common practice within the orthodontic office. Advertised advantages include providing a continuous gentle force, consistent long-lasting archwire seating, water sorption resistance and shape memory.[47] Realistically, the advantages include ease of use, patient comfort and the vast array of colors available. Readily seen disadvantages with elastomeric ligation include microbial accumulation, incomplete wire seating leading to poor tooth control, and binding of the bracket to the archwire.[47]

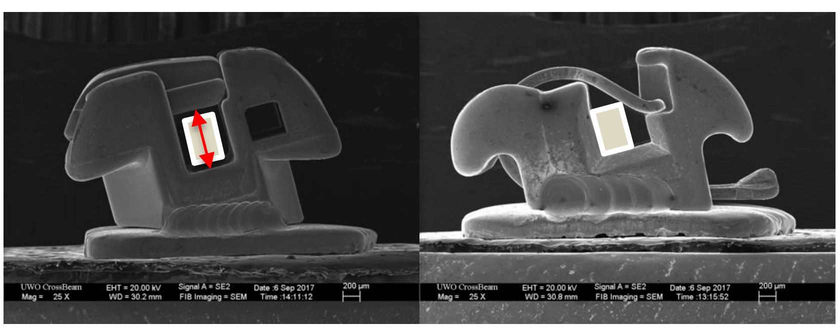
Elastomeric ligatures may not always provide the best ligation method for every scenario. They have been shown *in vitro* to lower torque values in an 0.022-in slot by approximately 20 percent when using a 0.019 x 0.025-in sized archwire, when compared to SS ligature ties.[31, 42, 47] This is largely due to the innate, rapid force decay that occurs as a result of the material composition and forces applied by the archwire. This rapid decay can be around 50 percent, even up to 70 percent, within the first 24 hours of use.[31, 42, 47] It is noted that this decay rate is an *in vitro* measurement and it is likely that the more extreme temperatures and acidity of the mouth may predispose the ligatures to even more rapid decay.[31]

Additionally, it has been documented that elastomers do not prevent bracket deformation to the same extent as SS ties when an archwire exerts axial moments on a twin bracket[1, 19], which can in turn lead to reduced torque expression. Although they seem less effective than SS ties, they are still used frequently due to their efficiency and ease of use. Often, the practitioner does not require the advantages that SS ties may offer, except in very specific scenarios.

1.4.2 Self-Ligation

The concept of self-ligation arose in the mid 1930's involving the Russell attachment, with the purpose of increasing clinical efficiency through quicker ligation times.[52] These brackets typically involve a door mechanism on the most buccal surface of the bracket that can be opened and closed. While the door is open, the archwire can be inserted into the bracket slot and the door closed afterwards to secure it in place. Since conception, there have been many purported advantages of self-ligating systems including increased patient comfort, better oral hygiene, increased patient cooperation, less chair time for clinicians and staff, a shorter treatment time, better patient acceptance of treatment and appliance, reduced friction, full and secure wire ligation, anchorage conservation, improved ergonomics and longer intervals between appointments.[16, 53] With the exception of reduced friction,[7] many of these claims have not been fully proven in any conventional study.

Within the broad spectrum of self-ligating bracket (SLB) systems, there exists "active" clips and "passive" doors (Figure 5 A and B). These terms relate to the door which may either have a spring clip mechanism, or rigid sliding piece. Active self-ligating brackets contain the spring clip mechanism which possesses the ability to press the wire into the bracket slot (Figure 5 B). In theory, this would help smaller wires express some degree of torque as they are able to engage the base of the slot more intimately during the earlier stages of treatment. This effect has been compared to conventional elastomers, with the potential for even more torque expression depending on the design of the spring clip. Further, the active spring clip potentially allows for greater rotational control, preventing slop in the first order.[7] Passive self-ligating brackets contain the rigid sliding door that transforms the bracket into a tube when closed (Figure 5 A). This has been proven to allow for very low friction mechanics during treatment[7], with little effect on increasing torque values.



(A) (B)

Figure 5: Examples of passive and active self-ligation systems: **A)** Passive self-ligation using a sliding door mechanism with slop between the wire and bracket; **B)** Active self-ligation using an active clip mechanism that engages the wire into the base of the bracket slot.

1.5 Archwires

The choice of archwire in orthodontic treatment can be complicated for many reasons. With the current technological advancements and access to materials, there is a vast array of sizes and compositions to address almost any concern during treatment. Archwire sizes typically range from very small 0.012-in round wires, up to very large 0.0215 x 0.028-in rectangular wires. These wires exhibit extremely different properties regardless of which material they are composed of. Additionally, changing their composition at any of these sizes will also have dramatic effects on properties such as range, springback, stiffness and strength.

1.5.1 Archwire Size

Archwire size plays a significant role in all aspects of orthodontic treatment and has a large influence on the amount of torque expression on the teeth. It is well known that different wire sizes possess uniquely different properties in terms of stiffness, strength and range.[8] With the ability to select between many different archwire sizes based on treatment progression, the orthodontist has many methods to alter the inclination of teeth in the final outcome.

A very generalized, typical wire sequence starts with smaller round wires that are not intended to express torque, but rather adapt to severe malocclusions for initial alignment and levelling. As treatment progresses, wire sizes increase to further fill the slot of the bracket and improve the general alignment of teeth. Near the final stages of treatment, large rectangular wires are often used which permit near-complete engagement of the bracket slot. As proven in the literature, increasing the cross-sectional diameter of these wires introduces significantly greater torque generation. As an example, changing from a 0.018 x 0.025-in to a 0.019 x 0.025-in nickel titanium wire provides approximately a 120% increase in torque.[18] These larger rectangular wires may further be twisted a certain number of degrees before insertion into the slot, creating plenty of axial moment on the bracket, which results in torque expression to the teeth.

Variations in the actual archwire cross-sectional diameters compared to the manufacturer's listed values has been of concern for many years. The wire's diameter has been well known to vary from that listed and is often smaller than the value given by the manufacturer.[4, 24–28] This directly decreases the amount of surface area engaged within the bracket slot, in turn reducing the ability to express torque. Accompanied by improper size, the wire edges are often bevelled, once again reducing their ability to engage the internal slot of the bracket as intimately as desired.[6, 14, 17, 25–27, 29–33] The smaller wire dimensions and bevelled edges are likely incorporated in an effort to make the insertion of large wires easier for the clinician. In theory, this edge-bevelling would also reduce the stiffness of every wire through a decrease in total wire material volume. On average, a total archwire volume loss of just below 8% can be expected, which could potentially reduce the stiffness of a wire up to 19% when compared to a wire that had perfectly squared edges.[29]

1.5.2 Archwire Composition

Archwires are typically manufactured with a handful of specific compositions to meet the needs of orthodontic treatment. The most common three wire types include nickel titanium (NT), titanium molybdenum alloy (TMA) and stainless steel (SS). Other wire compositions such as cobalt-chromium exist but have become outdated in contemporary treatment for most clinicians, as their properties are often inferior to the main three compositions.

Frequently, treatment will begin with using the small elastic NT wires that are readily adaptable and exceptional at aligning the dentition due to high range and springback properties. As treatment progresses, very exact movements of the teeth are required, leading to the introduction of TMA or SS wires. These stiffer archwires allow for precise finishing movements, such as torque. Historically, SS wires were the composition of choice during these final stages of treatment, but after the introduction of TMA around 1980, it wasn't long before the orthodontic community was experimenting with both types.[9] The traditional SS wires are very rigid and produce large forces over a relatively small range of action. TMA wires are very resilient and unique in their composition in that they are able to provide more gentle forces over a larger range of action.

The plastic yielding and strain hardening of these wires will differ from one to another as well and affect torque generation. It is known that a SS wire with a 0.019 x 0.025-in cross section has approximately three times the stiffness of the corresponding sized TMA wire.[8] The corresponding NT wire of 0.019 x 0.025-in size has four times less stiffness than the SS wire of this size.[33, 42] A study comparing the torque capability of the three main wire types found that in general the SS wires produced anywhere from approximately 1.5 – 2.0 times the torque of TMA wires, and approximately 2.5 – 3.0 times the torque of nickel titanium wires at angles of twist greater than 12 degrees.[15] Interestingly, at 12 degrees of twist or less, all three wires performed similarly, with no statistically significant differences. It may be that at such low angles of twist, the slot is only partially filled by the wires and the excessive slop in the bracket-archwire complex allows for only minimal torque expression, regardless of the wire type.[15]

The drastic change in torque produced by different wire compositions and sizes can easily be realized when comparing smaller nickel titanium to larger stainless steel wires. At similar torquing angles, using a 0.019 x 0.025-in stainless steel wire instead of an 0.018 x 0.025-in nickel titanium wire can provide up to 600% increased torque on an individual tooth.[18]

1.6 Torque Values

In order to perform these third-order movements of teeth, it is crucial to apply specific levels of torque in order to promote biologically safe moments on the dentition and the surrounding structures. Since the early days of orthodontics, torque values suggested for healthy movement of teeth have been quoted between 5 to 20 Nmm.[1, 4–7, 12–20] The lower value of 5 Nmm has been referred to as the minimum amount required for clinically significant torque of a maxillary incisor.[4, 17, 19, 21] Although these values are generally agreed upon in the existing literature, they are only suggested values. Of importance to note, however, is that higher forces for longer durations than normal can predispose teeth to more root resorption.[22, 23] For example, it has been found in recent experimental data that magnitudes of torque greater than 5 Nmm can cause more root resorption than normally seen in an untreated person.[22, 23]

1.6.1 Torque in Orthodontics

In orthodontics, the torque of teeth plays a critical role in both esthetics and occlusion.[1–3] The third-order movements of posterior teeth allow adequate coupling of the premolars and molars in a cusp-to-fossa relationship for proper function and adequate longevity of the dentition and accompanying periodontal structures. Detrimental tooth-to-tooth contacts that are premature or heavy can be a direct result of buccal or lingual crown tipping due to incorrect torque expression or bracket prescriptions on the teeth. The buccal-lingual inclinations of anterior teeth also play a critical role in the mouth. Proper torque of incisors allows for adequate overbite and overjet, along with alignment of the anterior teeth which is a significant factor in smile esthetics.[1, 2] Achieving ideal torque on the canine teeth allows for the fulfillment of canine guidance, which has long been held as an occlusal pattern that prevents harmful tooth wear and temporomandibular joint deterioration.[2, 3]

1.6.2 Torque Factors

Many factors have been studied and proven to significantly affect the values of torque on a tooth with regard to the brackets and archwires used during orthodontic treatment. The brackets can be assumed to have larger actual values[1, 4–7, 24, 26, 28, 30, 31, 34], and the wires to have smaller actual values[4, 24–28] when compared to the nominal values given from the manufacturer. Further, the contribution of edge bevelling on the wires reduces the intimate engagement of the wire and bracket slot even more.[6, 14, 17, 25–27, 29–31] Additionally, the wire stiffness can play a very significant role in the amount of torque expression within the bracket-archwire complex.[6, 14, 17, 25–27, 31]

When looking at the entire bracket-archwire complex, deformation of the system must be taken into account. Deformation of the wire during routine treatment will affect the wires ability to exert the desired force on the bracket and resultant tooth.[1, 19, 20, 26] When torsional forces are applied to the wire within the slot, deformation occurs due to the geometry of the archwire and location of forces directed at the wire.[1] As may be assumed, torque expression is often substantially less during unloading movements of the wire, as plastic deformation has already previously occurred during the loading process.[19]

In addition to the wire, a similar observation of elastic and plastic deformation of the bracket occurs during treatment[14, 17, 19, 20] and is more commonly seen when stiffer archwires are used with more plastic brackets.[17, 35] There can be significant differences in the hardness of the wires and brackets used, which can lead to one or the other being a major source of plastic deformation upon larger moments.[26] Steel brackets received after being used in active orthodontic treatment demonstrate deformed and notched internal walls of the slot, along with changes in the bracket dimensions from this plastic deformation.[26, 31] Further, this deformation is well known to vary between different brackets due to differing designs of the appliances themselves.[36]

There are additional factors that are often not thought of but may still contribute to the overall torque expression in the bracket-archwire complex. Acids from bacteria in the local micro-environment can bring about significant changes on the surface of stainless steel brackets[31, 37] that may reduce the engagement of wires. The incorrect placement of brackets, according to the manufacturers recommendations, has also proven to change the amount of torque exerted on teeth.[3, 6, 17, 38–43] In addition, the existing inclination of teeth, differing tooth morphology, and thickness of adhesive on the bracket base will also change torque expression.[5, 6, 13, 14, 17, 18, 24, 32, 38, 43–45] It is not hard to see that these could all contribute to the engagement angle of the archwire and bracket slot, which is yet another factor affecting how much torque expression is observed.[7, 24] Last, but not least, many studies have investigated and proven that the method of ligation between the bracket and archwire can change the torque expression values.[5, 14, 17, 31, 42, 46, 47]

1.7 Methods of Measuring Torque

In the recent literature, there are two common methods for measuring torque generated by a bracket-archwire complex. One method involves using an inclinometer to rotate a wire while it is ligated to a bracket on a rigid metal base. Using a multi-axis force transducer, the torque values exerted onto the rigid metal base from the archwire and bracket can be accurately measured.[1, 6, 13, 15, 19] This method is very technical and leaves little room for error if the bracket and archwire have not been aligned precisely. In order to center the wire perfectly in the bracket, it is possible to use very sensitive multi-dimensional load cells to detect even fractions of a Nmm in all three planes of space.[1, 13, 15, 19] The results from using a system like this are considered to be very accurate and specific to individual torquing of teeth without external stimuli that may affect the archwire.

Another method of testing torque values involves an orthodontic measuring and simulation system (OMSS). Using acrylic resin models, the largest wire possible is shaped to the appropriate archform and used to passively bond brackets onto the models. A force transducer that can measure forces and moments in all three planes of space is substituted for the tooth being evaluated. Torque may be applied at this point and recorded by the force transducer. Although not as technical, this method is seen quite frequently in the literature as well.[5, 17, 42, 49, 56] It is important to note that in this system, adjacent teeth in the OMSS model may give additional play to the archwire, changing the results depending on the specific model used.[4, 42]

Regardless of the method used to study torque, there are some factors that must be considered when interpreting the results of these studies. Ideally, it is important to understand how the bracket behaves in both directions of wire rotation. In studies that mount brackets using large rectangular wires, measuring both directions of wire rotation allows for more accurate depiction of the amount of slop on either side of the wire's starting position.

Another factor to consider when interpreting the results of torque studies is the position of the load cell relative to the bracket. When the load cell is attached directly to the bracket, torque is measured as forces are exerted on the buccal surface of the crown. Some studies have used mathematical transformations in order to record the torque at the bracket slot instead of on the load cell, although this is not common practice in the majority of the literature.[20] Another method would be to measure the torque values at a distance equal to the estimated center of resistance for a typical central incisor tooth. In this case, the load cell would be positioned approximately 10 mm from the bracket-archwire complex.[17] To avoid these issues, the load cell can be placed in such a way to record torque values exerted on the wires instead.

1.8 Previous Literature Investigating Torque

Studies previously conducted have attempted to address questions such as torque expression between bracket systems, torque expression between wire sizes and compositions, along with the engagement angle or slop of different bracket systems. Although not many in-depth studies are available in the literature regarding these topics, a few trends can be seen regarding wire rotation and torque generation for conventional SS working wires.

Possibly due to differences in methodology, it is currently unclear as to the exact behaviour of a bracket systems' ability to produce effective torque. A recently conducted study found that below 25 degrees of wire twist (or 35 Nmm of torque), there are essentially no statistically significant differences between bracket systems regarding their ability to produce torque.[19] In another study using an OMSS set-up, 6 different bracket systems subjected to 20 degrees of wire twist showed significantly different results from one bracket to another.[17] However, the methodology for this study is less clear and is subject to much more scrutiny. A systematic review addressing this same question found that clinically effective torque in the 5 – 20 Nmm range can be achieved around 15 – 31 degrees for active SLB's and around 22.5 – 34.5 degrees for passive SLB's.[14]

Comparing values of torque play or slop between bracket systems also seems to vary between studies. A study using a custom set-up revealed that active SLB's generally had less play than passive SLB's when compared to one another.[13] This same study found that these active SLB's tend to engage the wire sooner with higher torque values up to around 35 degrees of wire twist, then the values tended to equalize between systems. They concluded that active SLB's therefore had earlier engagement, but a wider variation of torque expression. Another study using the same set-up found that the slop was very similar between one passive SLB system and two active SLB's.[19] All three bracket systems were within one degree of each other, and the average torque play was close to 11 degrees. They concluded the slop was nearly indistinguishable between systems. However, one systematic review found that active SLB's tend to have engagement around 7.5 degrees whereas passive SLB's tend to have engagement around 14 degrees.[14]

A large portion of the previous literature falls short in providing data for different directions of wire twist. Rotation direction might have a significant effect on torque generated, especially in active SLB systems with varying clip designs. Although not many of the investigations are recent, those available tend to focus on specific values of wire rotation, without any mention of direction. Unfortunately, the few studies that do mention direction of wire twist differ greatly in methodology, making it hard to compare results.

One study by Sifakakis et al[42] reported values only at 15 degrees of wire twist in the clockwise and counterclockwise directions using a traditional twin bracket system.

This was the only study available that aimed to observe any differences between wire rotation direction. A study by Major et al[19] reported values for every 3 degrees from -15 up to 63 degrees. However, the direction of wire twist is not mentioned, and the methodology was designed to observe differences in loading and unloading curves as opposed to the differences in peak torque generation between wire twist directions. Another study by Major et al[20] reported values for loading and unloading curves at 16, 20, 24, 28, 32 and 40 degrees. Unfortunately, the focus of the investigation was on bracket deformation and not on differences in direction of wire rotation.

1.9 Summary of Issues

The literature available looking at torque generated with SS wires is quite limited. Previous studies have addressed the effects of large SS wires in only a handful of commonly used brackets within North America. There are no studies available comparing active and passive SLB systems to a traditional twin bracket system. Additionally, previous studies did not pursue the investigation of wire rotation direction on torquing capability. It seems prudent that information on the torquing capabilities of these new bracket systems with different clip mechanisms is revealed, as active clips may behave differently during different directions of wire twist. The benefits of testing them and comparing them to traditional twin bracket systems would seem very beneficial to the practicing clinician.

Chapter 2

2 Objectives and Hypothesis

2.1 Purpose of Current Investigation

The purpose of the current investigation was to compare the torque values generated by a conventional stainless steel working archwire engaged in three contemporary active SLB bracket systems, a contemporary passive SLB system and a traditional twin bracket system with elastic ligatures. Additionally, these systems were tested with wire rotations in the clockwise and counterclockwise directions. This information will help clinicians understand the torque generated at varying degrees and directions of wire rotation that can be expected with a conventional orthodontic working archwire in tandem with some of the most popular bracket systems currently available.

2.2 Hypothesis

- The active SLB systems will express higher torque at less rotation when compared to the passive SLB and traditional twin bracket system with elastic ligatures, due to the archwire seating action of the integrated clip.
- The torque expressed between active SLB bracket systems will differ depending on the direction of wire rotation and their proprietary clip designs.

Chapter 3

3 Materials and Methods

3.1 Orthodontic Bracket Systems

A total of five bracket systems were tested in this study. All of the tested brackets were of the same nominal 0.022 x 0.028-in (0.56 x 0.71 mm) slot size and all but one of the systems was of the self-ligation type. For the twin bracket that did not use self-ligation, a recently manufactured common grey elastomeric ligature (Ref: 854-660, Lot: K49716; American Orthodontics) was used to secure the wire to the bracket. A complete list of the studied brackets can be found in Table 1. These included Damon Q, Speed, In-Ovation R, Empower 2 Active and Victory Series (Figure 6 A – E). The selection of these bracket systems was based on the well-known reputation of the manufacturers and their common use in North America. Additionally, some have been tested in previous literature and would allow comparison of results and methodology.[13, 19, 20] The twin bracket was chosen as a traditional control, and it can be expected that the internal dimensions and mode of ligation should be similar to most other twin brackets that are currently available. The bracket prescription for each system was that which is most commonly used for each system. However, the bracket prescription should have no bearing within the study as all of the brackets are placed on a custom mounting jig in order to zero the tip and torque before being tested.

Ligation system	Test Group	Bracket System	Bracket Manufacturer	Bracket Item Number	Nominal Slot Size (in)
Passive Self-Ligation	P-Dmn	Damon Q	Ormco	491-6460	0.022 x 0.028
Active Self-Ligation	A-Spd	Speed	Speed System Orthodontics	0344	0.022 x 0.028
	A-Ovn	In-Ovation R	GAC Dentsply Sirona	189-112-80	0.022 x 0.028
	A-Emp	Empower 2 Active	American Orthodontics	485-1117	0.022 x 0.028
Twin Bracket with Elastic Ligature	E-Vic	Victory Series	3M Unitek	017-876	0.022 x 0.028

Table 1: Complete list of brackets categorized by type of ligation**.**

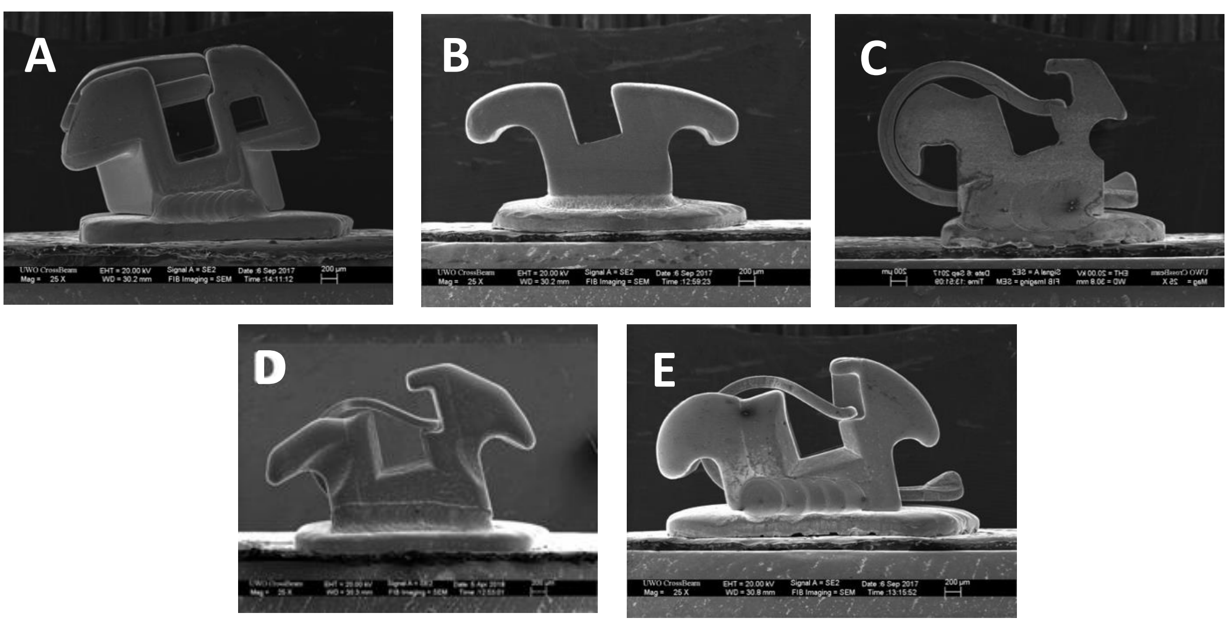
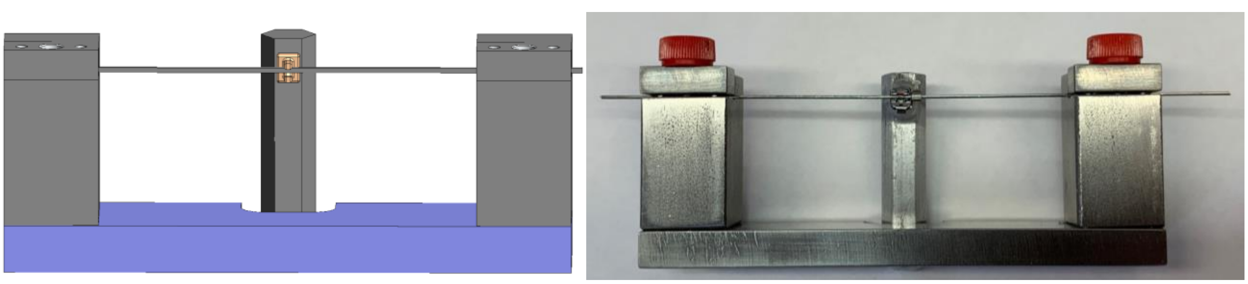


Figure 6: Scanning Electron Microscopy (SEM) imaging at 25x magnification of: **A)** P-Dmn; **B)** E-Vic; **C)** A-Spd; **D)** A-Ovn; **E)** A-Emp.

3.2 Custom Mounting Jig

A custom mounting jig was adapted from a previous study by Green et al[34] investigating different levels of friction within certain bracket systems (Figure 7). The entire jig was designed using computer aided design and manufacturing (CAD/CAM) and constructed from aluminum. It consisted of two rectangular poles on either end with clamps to center a 0.0215 x 0.028-in (0.55 x 0.71 mm) wire (Item # 03 125-58; GAC International) between them. Theoretically, a wire of this size should allow for close to zero slop within the slot of the bracket, completely negating the bracket prescription and allowing each bracket to be mounted in precisely the same position. A crimpable stop was placed on the wire just offset from the midline of the entire jig. The base segment had a small portion on one side that allowed the precise positioning of a six-sided hexagonal transfer pin in the middle of the jig. The transfer pins were also made out of rigid aluminum and could be securely mounted from a screw on the underside of the mounting jig. Once the mounting wire was secured in the jig, brackets could be ligated to the wire and slid flush against the stop, effectively centering them within the apparatus.

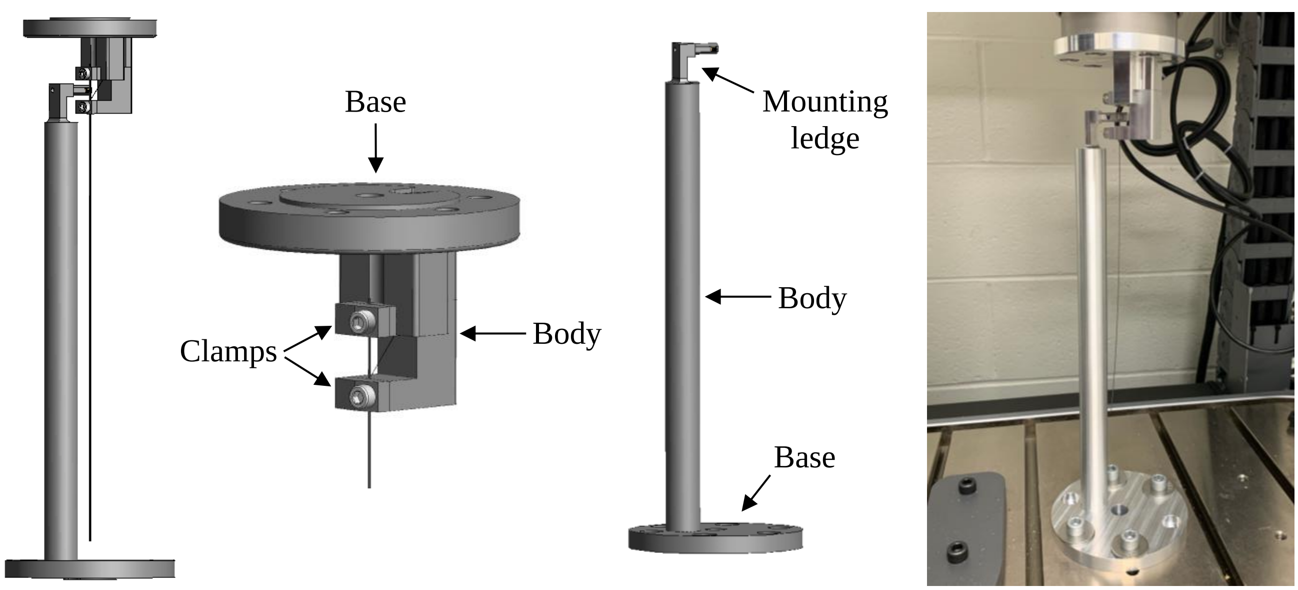


(A) (B)

Figure 7: Custom fabricated mounting jig from previous study with transfer mounting pin and mounting archwire: **A**) Computer rendered CAD/CAM model; **B)** Manufactured mounting jig with transfer pin.

3.3 Custom Torque Assembly

The custom torque assembly used in the current study was composed of two separate, but aligned parts. Both parts were fabricated from rigid aluminum using CAD/CAM with the help of Western University Machine Services (Figure 8). After meticulous measuring and fine-tuning, this assembly was designed with specifications to center a 0.019 x 0.025-in wire, along with a 0.022 x 0.028-in bracket slot down the exact center of an Instron materials testing machine (Instron Electropuls E10000; Norwood, MA, USA). Additional requirements included the ability to easily remove and replace the transfer pins, along with providing enough room for a long length of testing wire to facilitate efficient testing at each new section of wire.



(A) (B) (C) (D)

Figure 8: Custom torque assembly: **A)** Computer rendered CAD/CAM model of entire assembly; **B)** CAD/CAM model of clamp fixture mounted to the load cell on the Instron machine; **C)** CAD/CAM model of the base mounted to the baseplate of the Instron Machine with attachment for the transfer mounting pins; **D)** Actual manufactured entire custom torque assembly.

The base fixture, secured to the bottom of the Instron machine (Figure 8C), was designed with three distinct areas. A circular base was designed with six holes and a middle protrusion on the underside to positively seat in one specific location on the Instron Machine for every testing session. Six locating screws were used in the holes to rigidly secure the entire base fixture to the base of the Instron. The cylindrical body was long enough to hold the portion bearing the transfer pin high enough that a test wire of 14 inches (356 mm) in length could be easily accessible during testing when secured in place. The final section at the top of the base fixture has a rectangular portion that sits perpendicular to the body. A screw hole and small mounting ledge located on the most distal part of this section were designed to securely hold the transfer pins in the same position as the mounting jig. This allowed the internal dimensions of the bracket slot to be perfectly aligned with the central rotation axis of the Instron.

The clamp fixture secured to the load cell, subsequently connected to the actuator (Figure 8B), was also designed with three distinct regions. Similar to the base fixture, the circular base of the clamp fixture was designed to provide one repeatable position for mounting to the load cell. The rectangular body portion in the middle of the fixture was fabricated to allow the test wire to be slid upwards and trimmed between tests. Two separate clamps located on the most distal portion were designed with angled jaws that functioned to close precisely on a 0.019 x 0.025-in wire. Additionally, the clamps were precisely located to leave exactly 15 mm of wire length between each end to simulate the section of wire that would torque an upper right maxillary central incisor clinically. It was determined from measuring dentoform models, that 15 mm was approximately the average distance between the mesial edge of the upper right lateral incisor bracket and the mesial edge of the contralateral central incisor bracket (Figure 9).

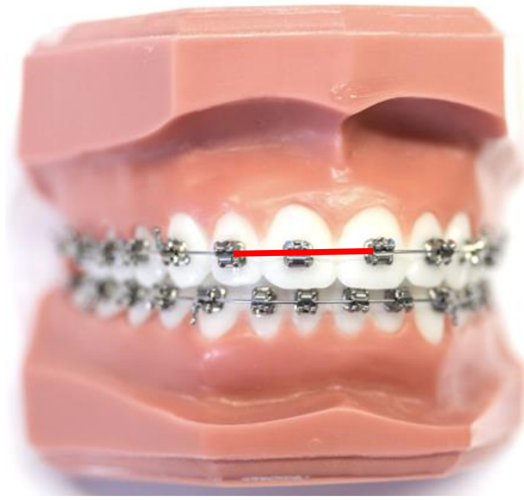


Figure 9: Dentoform model with 15 mm length of archwire highlighted in red.

3.4 Torque Testing

All of the brackets received from the manufacturer had their bases micro-etched with 50 micron aluminum oxide and wiped with alcohol, then mounted onto transfer pins using the custom mounting jig (Figure 7). The transfer pins were micro-etched in the same way as the brackets and single upper right central incisor brackets were placed on the transfer pins using Assure Plus primer (Ref: PLUS V7, Lot: 188218; Reliance Orthodontic Products) and Transbond XT light-cure adhesive (Item #: 0086, Lot: NA25697; 3M Unitek). Once aligned on the mounting jig, each bracket base was cured for three seconds on all four walls for a total of twelve seconds total per mounted bracket.

After mounting the brackets on the transfer pins, the pins were inserted into the custom torque assembly attached to the Instron materials testing machine (Figure 8). Once the brackets were secured within the custom torque assembly, a straight 0.019 x 0.025-in SS wire segment of 14 inches in length (Ref: 857-699SP; American Orthodontics) was inserted from the bottom into the wire slot within the assembly. On average, these wires measured with a digital caliper were 0.001-in smaller in both dimensions than the nominal values, which has been previously reported in the literature[24]. Next, the two angled screws were twisted to close the custom clamps on the wire, securing the 15 mm section of wire on either end of the bracket. The bracket door was then closed (or ligated with an elastomer in the case of E-Vic) and the assembly was ready for testing.

The Instron machine contained a 10 kN load cell (Serial #: 143580 {FORCE} and 143595 {TORQUE}) with a maximum torque capacity of +/- 100 Nm. This load cell was directly attached to the clamp portion of the custom torque assembly and as the machine rotated around the fixed base, the load cell recorded torque values using WaveMatrix2 Dynamic Testing software (Instron; Illinois Tool Works Inc. 2018; Norwood, MA, USA). Independent lab testing by Instron engineers with a set-up similar to the current study determined the error values with this specific test using the same load cell. A sample test torqued to the target value of 20.00 Nmm over 5 seconds and held at that value for 579 seconds showed a maximum jitter or noise of 9.10 Nmm (Figure 10). The average error value recorded over this 579 second timeframe when held at the target value of 20.00 Nmm was 2.86 Nmm.

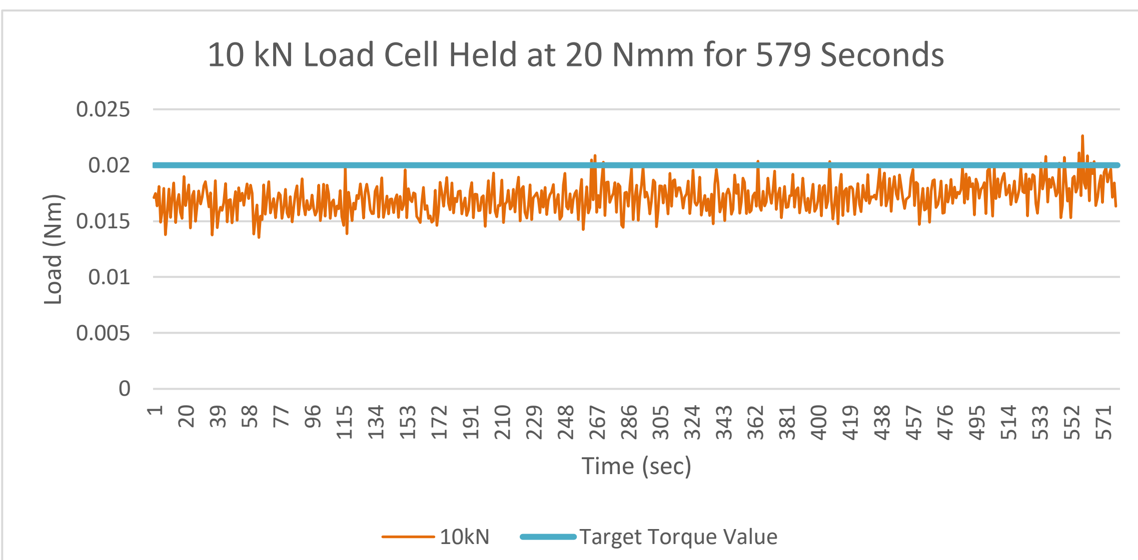


Figure 10: Independent lab testing of a sample held at the target value of 20 Nmm for 579 seconds.

Before testing, the wire and bracket slot were carefully centered visually, and it was confirmed that the wire was passively ligated into the bracket. After centering the bracket-archwire complex in this way, twenty brackets from each system were tested individually, ten in each direction. The first direction (termed clockwise) began with a counterclockwise wire rotation of 15 degrees, then a clockwise wire rotation of 60 degrees, and a final counterclockwise rotation back to the starting point of zero degrees. The second direction of tests (termed counterclockwise) were conducted in the same manner, but in the opposite direction with the wire rotated clockwise 15 degrees, then counterclockwise 60 degrees and finally clockwise back to the starting point. All wire rotations occurred at a rate of 1 degree per second and torque values were measured from negative 15 degrees to positive 45 degrees.

It was decided to rotate the wire in both the negative and positive directions during each test for a few reasons. The initial rotation of 15 degrees was performed in order to allow the bracket to engage the wire slightly in the direction that was not the main wire rotation direction for that test. Along with the data from the main direction of wire rotation, this would allow a comparison of the values of engagement on either side of the zero point, as well as determine the bracket “slop”. It would also allow an adjustment of values (if required) in order to ensure the wire was centered in the bracket slot. This served as a check to confirm that the starting points were equal for each group and the values between directional tests could be adequately compared and related to each other.

After each test, the Instron machine was stopped while the wire was moved within the assembly until an unaltered segment of wire was reached. The load cell was re-calibrated to zero Nmm of torque and the transfer pin was removed and rotated until the next bracket was aligned in the assembly. The used portion of wire was then trimmed from the top, the unused wire was secured once again, and the new bracket door was closed (or ligated) before a new test took place. One test was conducted for each bracket with a new segment of wire and new individual bracket each time.

3.5 Data Analysis

After data collection (Appendix A), an adjustment was performed for each individual bracket system in order to approximate the angle of engagement. Theoretically, the rotation angles in the counterclockwise direction for the clockwise tests should be the same angles as those in the clockwise direction for the counterclockwise wire rotation tests. Therefore, an adjustment was done by matching the same data points from the clockwise and counterclockwise rotations. For example, the P-Dmn clockwise data set showed a value of 3.5 Nmm at +4 degrees of wire twist and the counterclockwise data set showed a value of 3.5 Nmm at -9 degrees of wire twist (Appendix A). In order to approximate these points with each other, the entire set of P-Dmn clockwise data was adjusted by adding 5 degrees to each angle of rotation (Appendix C). This process was conducted for each bracket system within their own set of data points in order to confirm the angle of engagement (Appendix C – G) and ensure accurate comparison between bracket systems. The adjusted results are presented at every degree of rotation in Appendix B.

Descriptive statistics including mean and standard deviation (SD) were calculated for each bracket and rotation direction combination. A statistical software package (SPSS Statistics 24.0; SPSS, Inc., Chicago, IL) was used to carry out a two-way analysis of variance (ANOVA) with Bonferroni correction for multiple comparisons to compare significant differences between groups for every 3 degrees of rotation. Statistical significance was set at $P<0.05$.

Chapter 4

4 Results

The mean torque (±SD) generated at each degree of rotation for each of the five bracket systems is shown in Table 2. Data is reported at every 3 degrees of rotation, from -9 degrees to +42 degrees. Tests are divided into clockwise (“Clock”) and counterclockwise (“Counter”) rotation columns for each bracket system. The values from Table 2 are plotted in graphical format in Figures 11 and 12. Figure 13 shows the significant differences between groups ($P<0.05$) at every 3 degrees of rotation with error bars representing 1 SD.

Torque (Nmm)										
	Damon		Speed		InOvation-R		Empower		Victory	
Angle(°)	Clock	Counter	Clock	Counter	Clock	Counter	Clock	Counter	Clock	Counter
-9	3.8 (1.4)	3.5 (1.3)	0.9 (0.4)	1.5 (0.4)	1.2 (0.3)	3.8 (1.7)	3.0 (0.6)	2.4 (0.4)	1.8 (1.2)	2.5 (0.5)
-6	0.7 (0.9)	1.5 (1.0)	0.5 (0.3)	0.9 (0.3)	0.9 (0.4)	2.6 (0.6)	2.6 (0.7)	2.5 (0.5)	1.6 (0.7)	1.8 (0.4)
-3	0.4 (0.5)	1.0 (0.8)	0.4 (0.2)	0.7 (0.2)	0.9 (0.5)	2.2 (0.9)	1.5 (0.8)	2.1 (0.8)	1.3 (0.4)	1.4 (0.5)
0	0.6 (0.4)	0.5 (0.3)	0.3 (0.2)	0.4 (0.1)	0.7 (0.6)	0.8 (0.5)	0.8 (0.7)	1.8 (0.6)	0.5 (0.4)	0.7 (0.4)
3	0.8 (0.4)	0.4 (0.5)	0.3 (0.3)	0.6 (0.4)	2.2 (0.2)	0.9 (0.5)	2.6 (0.5)	2.3 (0.5)	1.9 (0.6)	1.3 (0.3)
6	2.0 (1.1)	0.7 (0.8)	1.0 (0.4)	0.7 (0.6)	2.2 (0.8)	1.2 (0.4)	3.0 (0.5)	2.8 (1.1)	2.3 (0.5)	1.6 (0.3)
9	3.5 (1.1)	3.2 (1.3)	1.3 (0.4)	1.0 (0.8)	3.4 (2.4)	1.3 (0.3)	2.7 (0.6)	3.5 (2.6)	2.3 (0.7)	2.1 (0.7)
12	8.6 (1.4)	9.2 (1.4)	3.7 (1.3)	2.6 (1.2)	9.0 (3.6)	3.5 (1.8)	2.9 (0.9)	5.9 (4.3)	5.3 (1.3)	5.4 (1.3)
15	16.1 (1.5)	16.4 (1.5)	9.3 (1.6)	6.9 (1.6)	16.9 (3.9)	9.1 (3.0)	5.2 (2.8)	11.2 (4.7)	13.1 (1.8)	11.5 (1.7)
18	24.4 (1.7)	23.3 (1.7)	15.5 (1.2)	12.9 (2.1)	25.5 (4.1)	16.1 (3.2)	9.8 (4.2)	17.6 (5.0)	22.4 (1.8)	18.8 (2.0)
21	33.1 (1.7)	30.7 (2.1)	21.8 (1.0)	19.8 (2.4)	34.5 (4.1)	23.3 (3.2)	16.3 (4.9)	24.3 (5.1)	32.2 (1.9)	25.9 (2.0)
24	42.6 (1.8)	39.1 (1.9)	27.2 (1.3)	26.8 (2.5)	43.3 (4.0)	30.6 (3.4)	23.2 (5.3)	31.1 (5.8)	42.5 (1.7)	33.3 (2.2)
27	51.8 (1.7)	47.8 (2.0)	32.4 (1.6)	34.9 (2.7)	52.0 (4.0)	39.4 (3.9)	30.3 (5.6)	39.0 (6.0)	52.3 (1.8)	42.2 (2.3)
30	60.8 (1.7)	56.1 (1.8)	37.5 (2.1)	43.4 (2.9)	60.5 (3.8)	48.4 (3.8)	37.4 (6.0)	47.7 (5.8)	61.6 (1.7)	51.1 (2.4)
33	69.2 (1.6)	64.6 (1.8)	42.8 (3.0)	52.1 (2.8)	68.2 (3.5)	57.1 (4.0)	44.5 (6.0)	56.6 (5.5)	70.4 (1.6)	60.0 (2.0)
36	76.9 (1.5)	72.7 (1.4)	48.1 (3.8)	60.7 (2.7)	75.5 (3.3)	66.3 (3.8)	51.3 (5.9)	65.5 (5.1)	78.3 (1.6)	68.7 (2.0)
39	83.9 (1.3)	80.2 (1.4)	52.9 (4.7)	69.1 (2.3)	82.0 (3.0)	74.8 (3.5)	57.9 (5.7)	73.3 (4.5)	85.4 (1.5)	76.9 (1.8)
42	89.8 (1.1)	86.8 (1.1)	57.2 (5.8)	76.8 (2.2)	87.9 (2.7)	82.3 (2.9)	63.9 (5.6)	80.6 (3.9)	91.2 (1.3)	84.1 (1.7)

Table 2: Mean torque (±SD) for each bracket system at every 3° of wire rotation in both clockwise and counterclockwise directions.

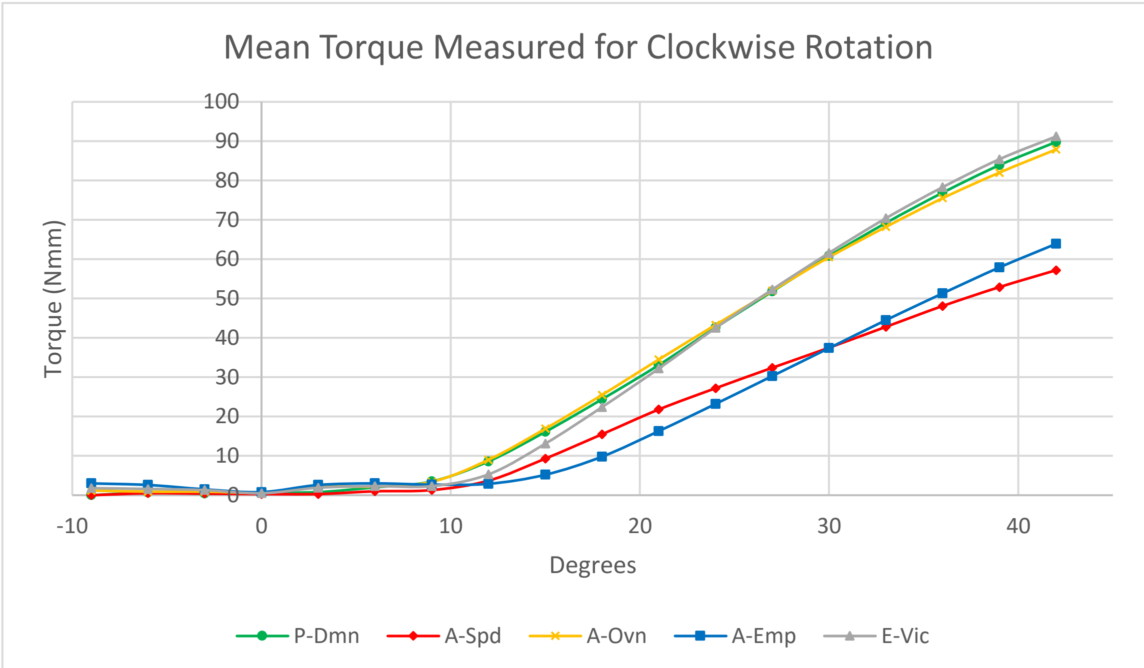


Figure 11: Mean torque values at every 3° of rotation for each bracket system in the clockwise direction.

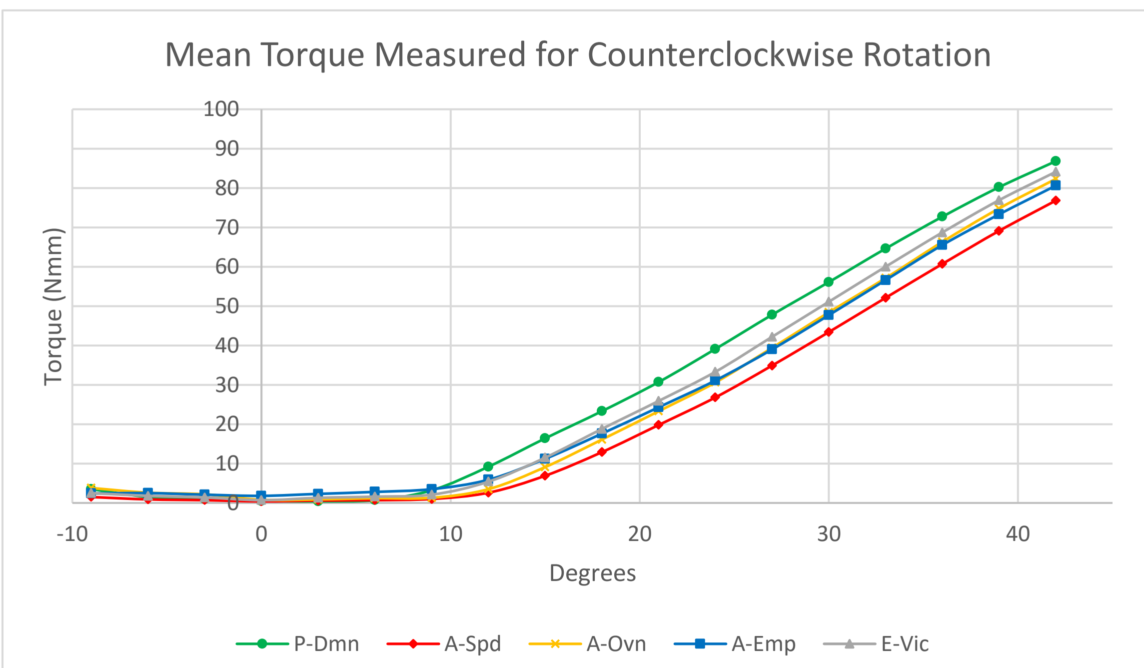
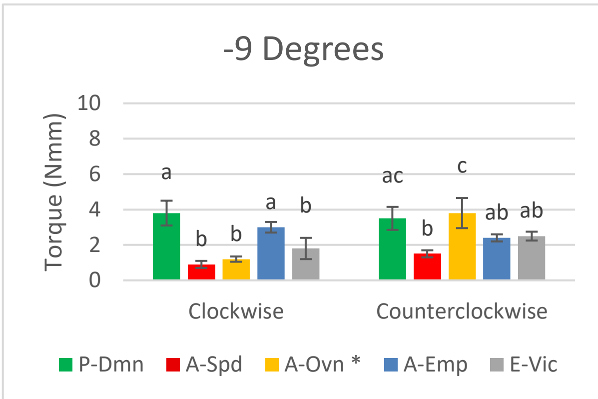
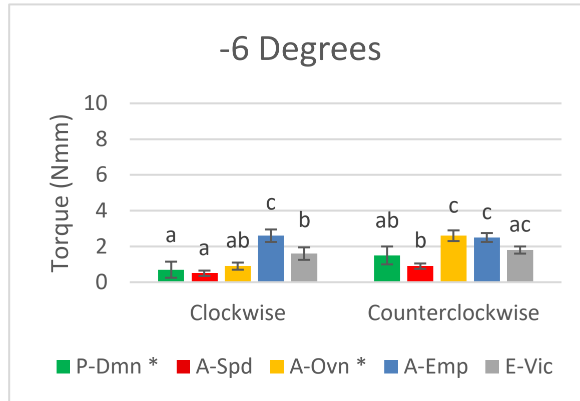


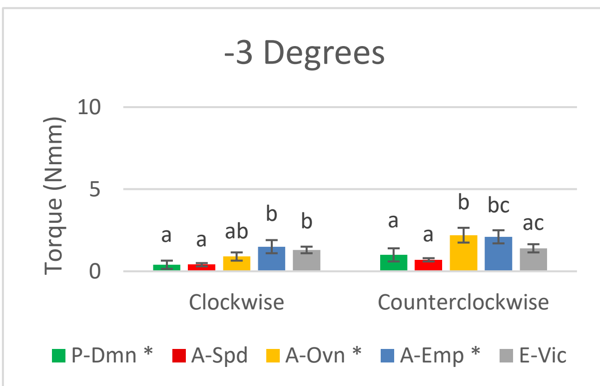
Figure 12: Mean torque values at every 3° of rotation for each bracket system in the counterclockwise direction.


-9 Degrees
Torque (Nmm)
10
8
6
4
2
0
a b b a b
ac b c ab ab
Clockwise
Counterclockwise
P-Dmn
A-Spd
A-Ovn *
A-Emp
E-Vic

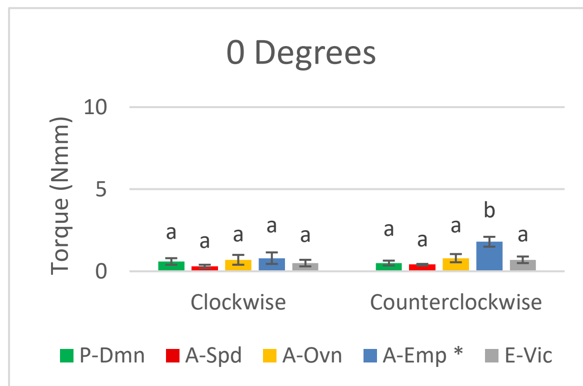
(A)


-6 Degrees
Torque (Nmm)
10
8
6
4
2
0
a a ab c b
ab b c c ac
Clockwise
Counterclockwise
P-Dmn *
A-Spd
A-Ovn *
A-Emp
E-Vic

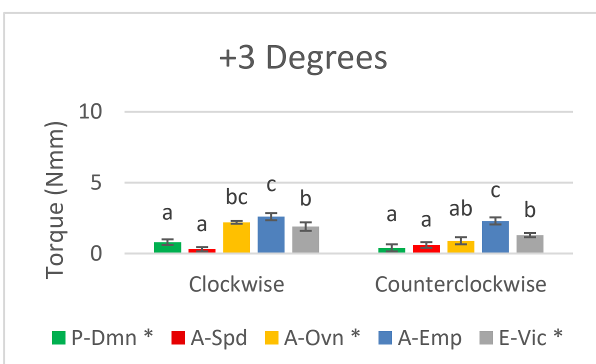
(B)


-3 Degrees
Torque (Nmm)
10
5
0
a a ab b b
a a b bc ac
Clockwise
Counterclockwise
P-Dmn *
A-Spd
A-Ovn *
A-Emp *
E-Vic

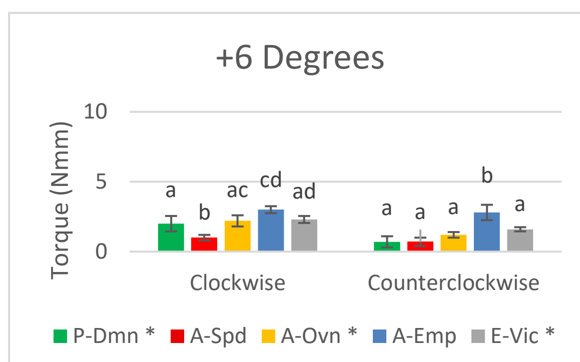
(C)


0 Degrees
Torque (Nmm)
10
5
0
a a a a a
a a a b a
Clockwise
Counterclockwise
P-Dmn
A-Spd
A-Ovn
A-Emp *
E-Vic

(D)


+3 Degrees
Torque (Nmm)
10
5
0
a a bc c b
a a ab c b
Clockwise
Counterclockwise
P-Dmn *
A-Spd
A-Ovn *
A-Emp
E-Vic *

(E)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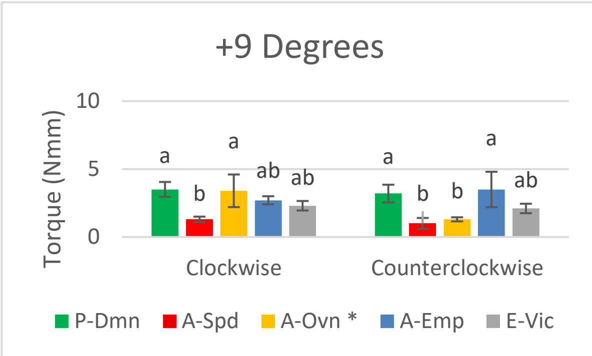
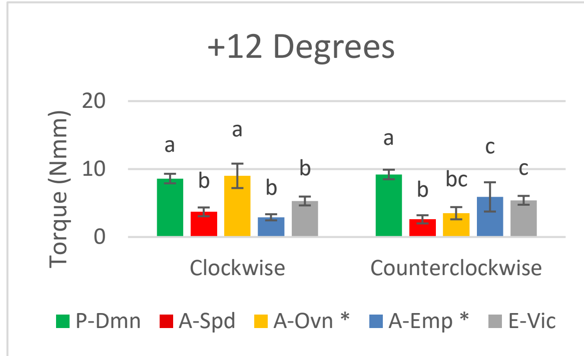
Torque (Nmm)
10
5
0
a b ac cd ad
a a a b a
Clockwise
Counterclockwise
P-Dmn *
A-Spd
A-Ovn *
A-Emp
E-Vic *

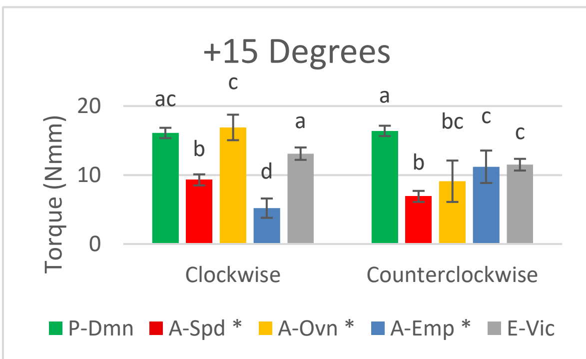
(F)


+9 Degrees
Torque (Nmm)
10
5
0
a b a ab ab
a b b a ab
Clockwise
Counterclockwise
P-Dmn
A-Spd
A-Ovn *
A-Emp
E-Vic

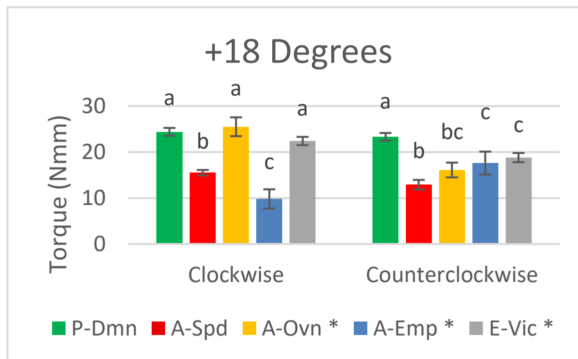
(G)


+12 Degrees
Torque (Nmm)
20
10
0
a b a b b
a b bc c c
Clockwise
Counterclockwise
P-Dmn
A-Spd
A-Ovn *
A-Emp *
E-Vic

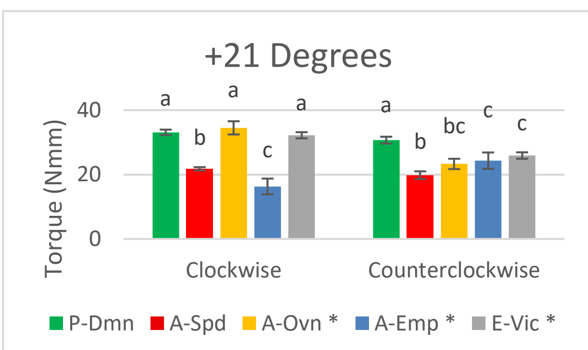
(H)


+15 Degrees
Torque (Nmm)
20
10
0
ac b c d a
a b bc c c
Clockwise
Counterclockwise
P-Dmn
A-Spd *
A-Ovn *
A-Emp *
E-Vic

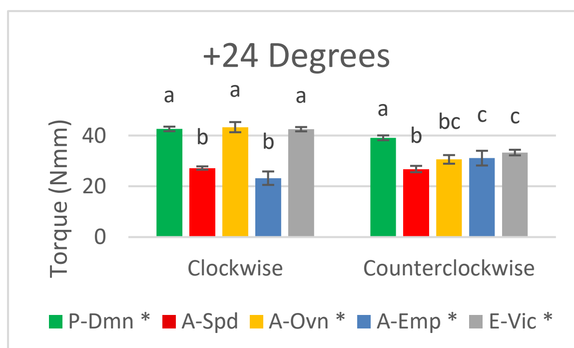
(I)


+18 Degrees
Torque (Nmm)
30
20
10
0
a b a c a
a b bc c c
Clockwise
Counterclockwise
P-Dmn
A-Spd
A-Ovn *
A-Emp *
E-Vic *

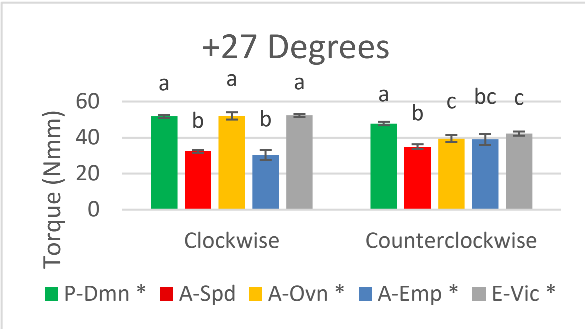
(J)

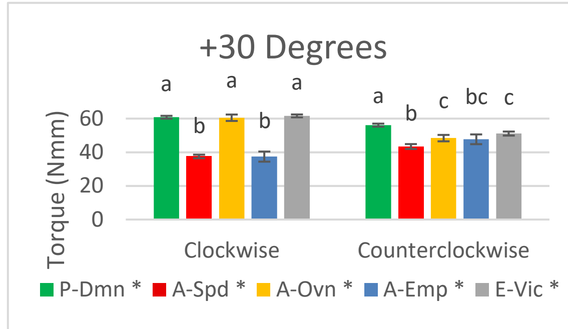

+21 Degrees
Torque (Nmm)
40
20
0
a b a c a
a b bc c c
Clockwise
Counterclockwise
P-Dmn
A-Spd
A-Ovn *
A-Emp *
E-Vic *

(K)


+24 Degrees
Torque (Nmm)
40
20
0
a b a b a
a b bc c c
Clockwise
Counterclockwise
P-Dmn *
A-Spd
A-Ovn *
A-Emp *
E-Vic *

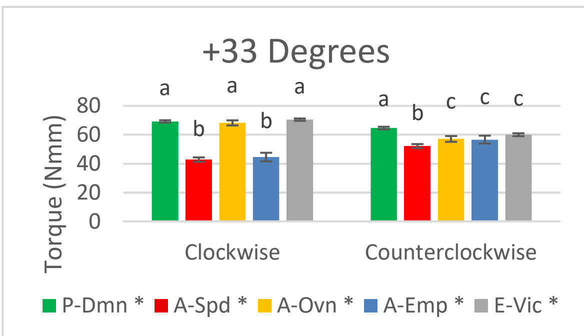
(L)

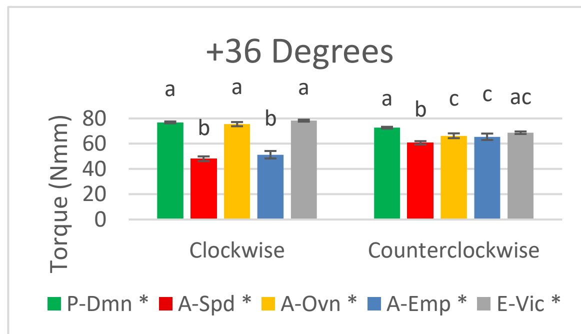




(M)

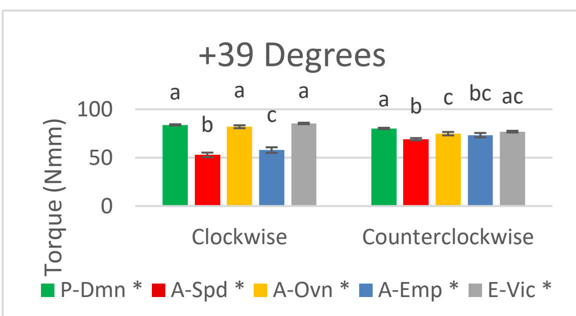
(N)

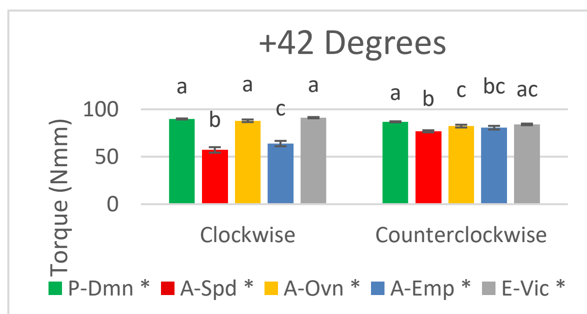




(O)

(P)





(Q)

(R)

Figure 13: Bar graphs showing mean torque values (Nmm) obtained for each bracket system in both clockwise and counterclockwise rotations. Similar letters denotes no statistically significant difference between groups in the same rotation direction. An "*" beside the bracket system group denotes a statistically significant difference in torque between clockwise and counterclockwise rotation directions.

4.1 Degrees of Rotation

In the clockwise rotation tests, the torque values generated at the maximum +42 degrees of rotation for P-Dmn, A-Spd, A-Ovn, A-Emp and E-Vic are 89.8, 57.2, 87.9, 63.9 and 91.2 Nmm respectively. The torque values for P-Dmn, A-Ovn and E-Vic were significantly higher than those recorded for A-Spd and A-Emp ($P<0.001$; ES=0.887; Power=1.000) (Figure 13R).

In the counterclockwise rotation tests, the maximum torque values at +42 degrees for P-Dmn, A-Spd, A-Ovn, A-Emp and E-Vic are 86.8, 76.8, 82.3, 80.6 and 84.1 Nmm respectively. The values for P-Dmn and E-Vic were the highest and significantly different than A-Spd ($P<0.001$; ES=0.887; Power=1.000) (Figure 13R). Interestingly, all of the systems generated torque values within 10 Nmm of each other in this direction of rotation.

Other angles of clinical importance include +12 and +24 degrees, as these angles include the range of common bracket prescriptions used during treatment.[2, 3] The torque values for P-Dmn, A-Spd, A-Ovn, A-Emp and E-Vic at +12 degrees in the clockwise direction were 8.6, 3.7, 9.0, 2.9 and 5.3 Nmm, respectively. The values for P-Dmn and A-Ovn are significantly higher than those of A-Spd, A-Emp and E-Vic ($P<0.05$; ES=0.478; Power=1.000) (Figure 13H).

The torque values for P-Dmn, A-Spd, A-Ovn, A-Emp and E-Vic at +12 degrees in the counterclockwise direction are 9.2, 2.6, 3.5, 5.9 and 5.4 Nmm, respectively. P-Dmn generated a significantly higher torque than all the other systems at this angle ($P<0.01$; ES=0.478; Power=1.000)(Figure 13H). A-Spd and A-Ovn generated the lowest torque values at this early stage of counterclockwise rotation.

At +24 degrees of rotation in the clockwise direction, the values for P-Dmn, A-Spd, A-Ovn, A-Emp and E-Vic were 42.6, 27.2, 43.3, 23.2 and 42.5 Nmm, respectively. Similar to the trend seen at +42 degrees, the values for P-Dmn, A-Ovn and E-Vic are significantly higher than those of A-Spd and A-Emp ($P<0.001$; ES=0.769; Power=1.000) (Figure 13L).

The values for P-Dmn, A-Spd, A-Ovn, A-Emp and E-Vic at +24 degrees in the counterclockwise direction were 39.1, 26.8, 30.6, 31.1 and 33.3 Nmm, respectively. P-Dmn again had significantly higher torque generation than the other systems ($P<0.001$; ES=0.769; Power=1.000)(Figure 13L). A-Spd generated the lowest value, although the difference between other systems is much smaller in this direction.

4.2 Direction of Rotation

In general, bracket systems performed in the same manner, with similar slopes, when tested in the counterclockwise direction. However, when tested in the clockwise direction, the A-Spd and A-Emp brackets performed quite differently from the other brackets and compared to their respective counterclockwise behaviour.

For the P-Dmn brackets, significant differences in wire rotation direction were noted at -6, -3, +3, +6 and from +24 up until the final +42 degrees ($P<0.05$) (Figure 13 and 14) with more torque being generated in the clockwise rotation group. The absolute difference peaked at 4.7 Nmm with +30 degrees of rotation.

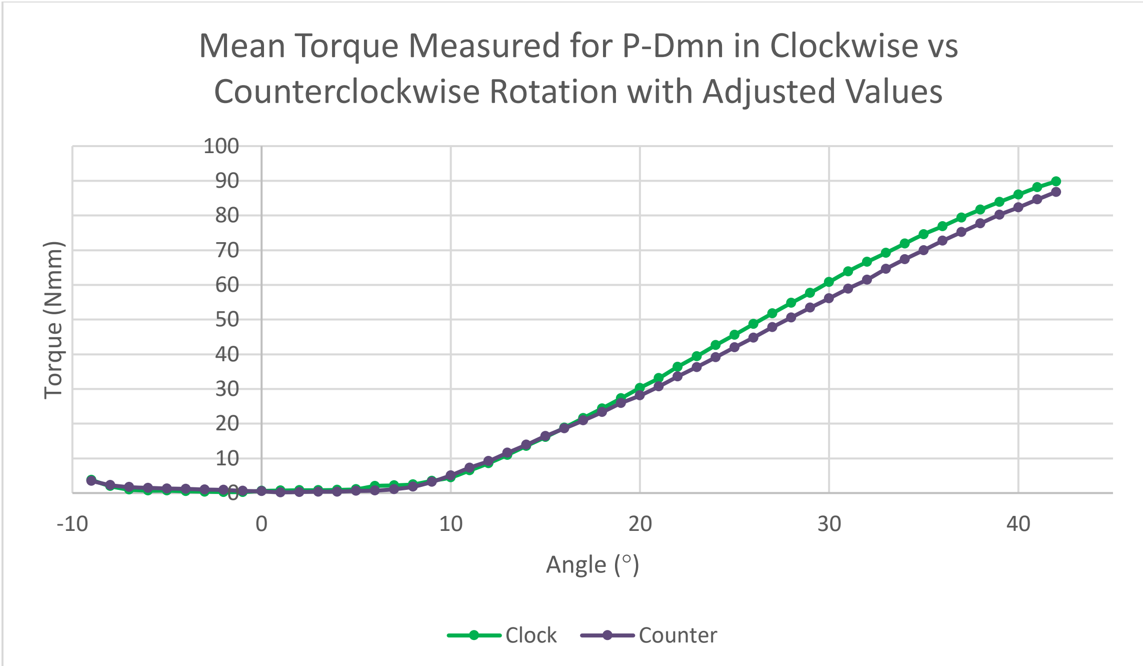


Figure 14: Mean P-Dmn torque values from -9 to +42 degrees in clockwise and counterclockwise rotation tests.

The results for the E-Vic brackets presented in a similar manner. Significant differences in wire rotation direction were noted at +3, +6 and then from +18 up until the final +42 degrees ($P<0.05$) (Figure 13 and 15), with more torque being generated during clockwise rotation. The difference reaches a maximum of 10.5 Nmm at +30 degrees.

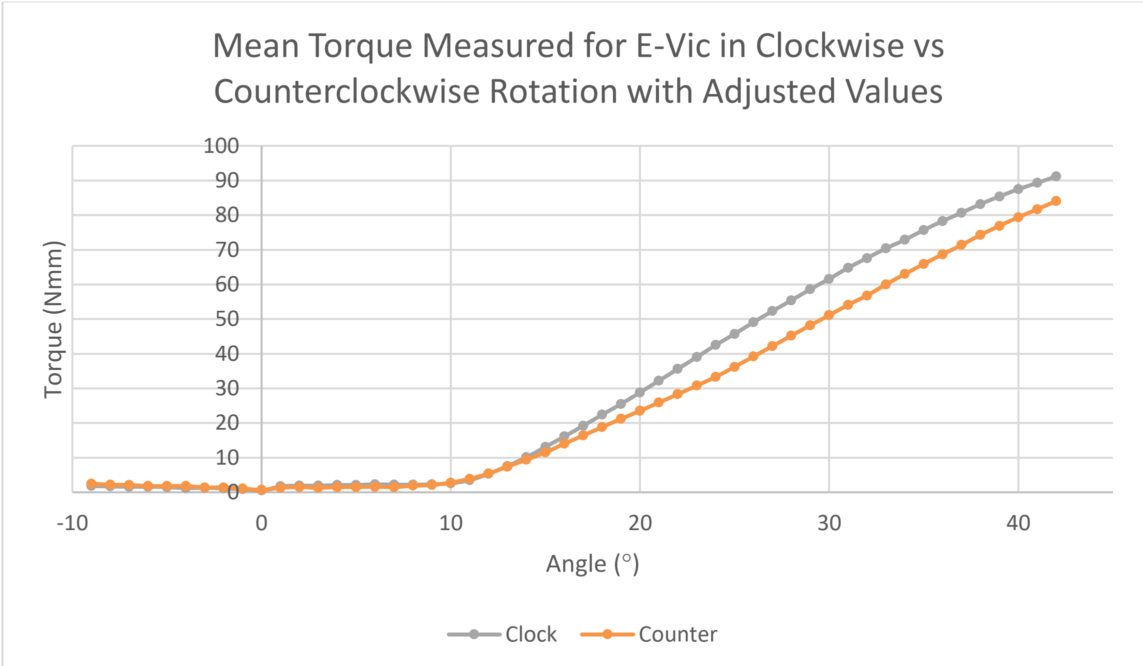


Figure 15: Mean E-Vic torque values from -9 to +42 degrees in clockwise and counterclockwise rotation tests.

For the A-Ovn brackets, significant differences in wire twist direction were noted at every degree of rotation except at zero degrees ($P<0.005$) (Figure 13 and 16), with more torque generated consistently in the clockwise direction. At +24 degrees, the differences peak at a maximum of 12.7 Nmm.

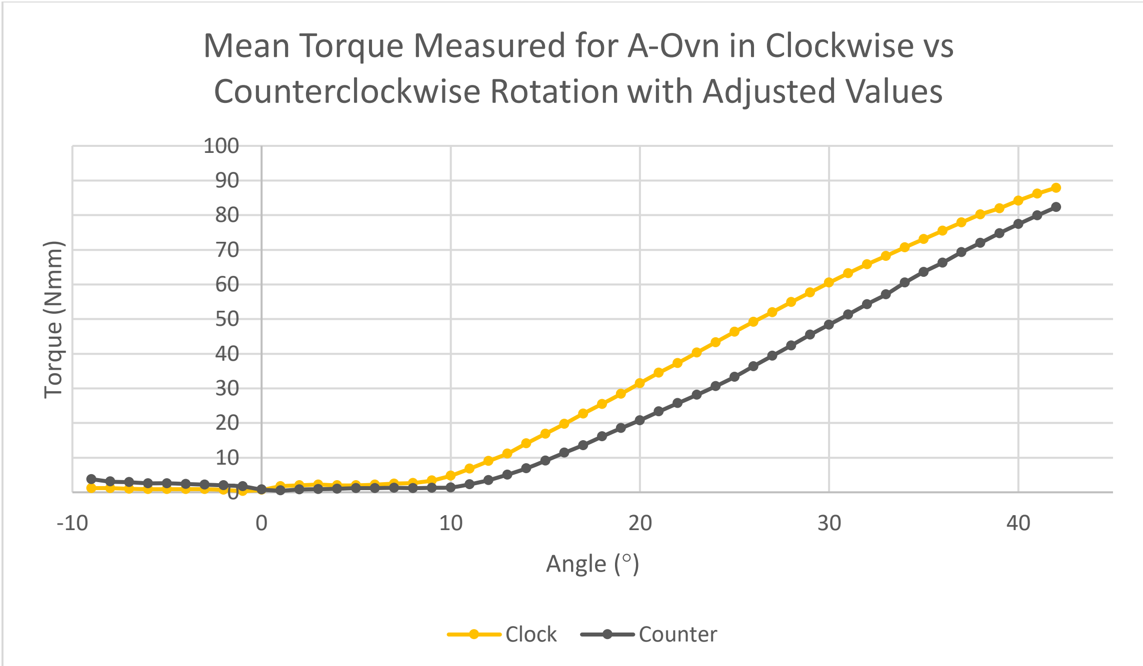


Figure 16: Mean A-Ovn torque values from -9 to +42 degrees in clockwise and counterclockwise rotation tests.

For the A-Spd bracket, significant differences in wire rotation direction were noted at +15 and then from +30 up to the final +42 degrees ($P<0.05$) (Figure 13 and 17) but the direction that generated higher torques was opposite to the P-Dmn, E-Vic and A-Ovn bracket systems. This difference between directions climbs up to 19.6 Nmm at +42 degrees, with more torque generation in the counterclockwise rotation group.

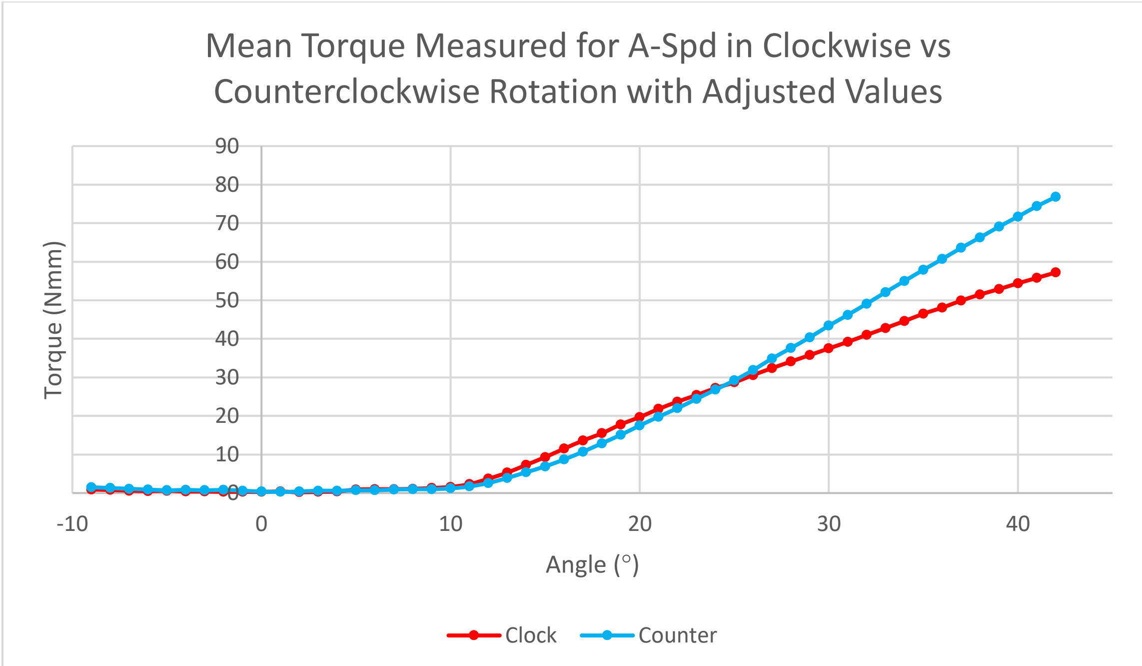


Figure 17: Mean A-Spd torque values from -9 to +42 degrees in clockwise and counterclockwise rotation tests.

The results for A-Emp were similar to those of A-Spd and opposite to those of the other bracket systems, with more torque generated in the counterclockwise direction. Statistically significant differences are noted at -3, 0 and then from +12 up until +42 degrees of twist ($P<0.05$) (Figure 13 and 18). The differences at +12 up until +42 degrees continue to increase in magnitude as the amount of wire twist increases, up to a maximum difference of 16.7 Nmm at +42 degrees of rotation.

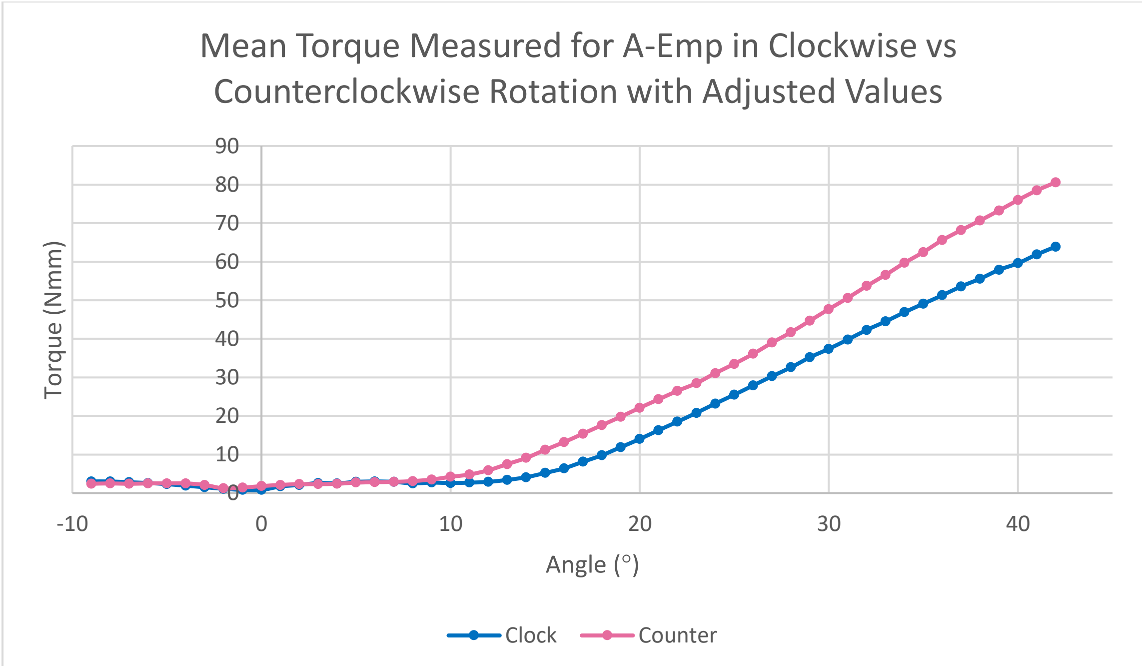


Figure 18: Mean A-Emp torque values from -9 to +42 degrees in clockwise and counterclockwise rotation tests.

4.3 Torque Play and Angles of Engagement

To determine the amount of torque play within each bracket system, visual extrapolation was performed on the mean values graph for each direction of wire rotation (Figure 11 and 12). Angles were chosen where it appeared the slope of the graph changed by more than 10% in magnitude and were rounded to the nearest degree. The values for engagement on either side of the zero point that were extrapolated using this method are reported in Table 3.

	Clock Engagement(°)	Counter Engagement(°)	Average Engagement(°)	Average Range of Torque Play (°)
P-Dmn	-7, +8	-8, +8	-7.5, +8	15.5
A-Spd	-10, +10	-11, +10	-10.5, +10	20.5
A-Ovn	-10, +8	-8, +10	-9, +9	18
A-Emp	-10, +12	-11, +10	-10.5, +11	21.5
E-Vic	-9, +10	-9, +9	-9, +9.5	18.5

Table 3: Average angles of engagement and torque play for each bracket system.

Chapter 5

5 Discussion

The aim of this study was to compare torque expression using a conventional SS working wire between three contemporary active SLB systems, one contemporary passive SLB and a traditional twin bracket system as a control. Unlike previous studies, both directions of wire rotation (clockwise and counterclockwise) were performed to determine if the bracket systems would behave in different manners from one direction to the next. A novel method of measuring torque in a custom device using an Instron machine was devised to perform the tests. With this unique methodology and the wide selection of brackets used, this investigation hopes to reveal important information regarding the torquing ability of contemporary SLB's.

5.1 Methodology to Study Torque

This study was performed using a novel, custom designed, experimental set-up. The custom mounting jig and torquing assembly were designed to precisely position the slot of an 0.022-in bracket and a 0.019 x 0.025-in SS wire down the central rotation axis of the Instron. The methodology used in this investigation can be directly compared to a small selection of other studies.[13, 15, 19-20] All four of these other comparable studies used a custom orthodontic torque measurement device with a worm gear at a ratio of 1:120 and a multi-axis force/torque transducer. The gear ratio of 1:120 allowed for measurements at every 3 degrees of rotation. The current investigation aimed to improve on this by using a device that allowed for a high level of precision and versatility for recording and adjusting rotation to every thousandth of a degree at a specific rate. This combination of the Instron machine and software allowed for consistent and reliable testing methods between each individual bracket and for the different directions of rotation.

The 10 kN load cell with a 100 Nm maximum torque capacity used on the Instron machine was confirmed to have a total range of torque values of 6.63 Nmm and an average error value of 2.86 Nmm, when torqued to the target value of 20.00 Nmm for 579 seconds. Unique to this investigation only, these values were recorded and reported by independent lab testing from Instron using a similar set-up. Reporting the accuracy of the load cell to this degree was very important as it allows for more certainty in the validity of the results. No other previous study was found to provide accuracy data such as this, so directly comparing equipment to other literature was not possible. Although this load cell data provides a large improvement over previous studies, the load cell itself is likely limited in its accuracy at low angles of rotation that generate very small values of torque.

In this study, the wire and bracket slot were visually centered as best as possible, and the mounting jig contained a 0.0215 x 0.028-in wire in order to negate most of the slop. Additionally, the Instron machine was able to closely reproduce the starting position for each individual bracket throughout testing. However, some energy was inevitably lost in the system potentially leading to minute alterations in starting points over time. This hysteresis can be noted between the starting and ending points in Appendix H. In order to accommodate for these minor differences, the data was adjusted within each bracket system and direction of rotation before completing the data analysis. A thorough range of values from -9 to +42 degrees of rotation were available for analysis and allowed direct comparison with previous studies.

5.2 Degrees of Wire Rotation

This study was novel in the approach of measuring torque with an Instron machine from -15 to +45 degrees of wire rotation in both the clockwise and counterclockwise directions. A study done by Major et al.[19] was conducted in a similar fashion to the current investigation. Although only one direction of wire rotation was tested (it was not stated in which direction), three of the same bracket systems were used in their study (P-Dmn, A-Spd and A-Ovn). The values they obtained at +42 degrees of wire twist were 81.5, 70.4 and 86.3 Nmm for their P-Dmn, A-Spd and A-Ovn, respectively. This is similar to the counterclockwise values obtained in this current investigation of 86.8, 76.8 and 82.3 Nmm for the P-Dmn, A-Spd and A-Ovn, respectively for the same bracket systems. Peak torque values are comparable between studies and the small differences may be a function of minor variations in methodology.

At angles of +12 degrees in the Major et al.[19] study, they found values of 4.6, 3.6 and 3.8 Nmm for their P-Dmn, A-Spd and A-Ovn, respectively. Again, these are most comparable to the counterclockwise values found in the current investigation of 9.2, 2.6 and 3.5 Nmm, for the same bracket systems. Interestingly, the results in the current study for Damon Q were higher in both directions of twist.

At angles of +24 degrees in the Major et al.[19] study, they found values of 33.4, 29.7 and 33.8 Nmm for their P-Dmn, A-Spd and A-Ovn, respectively. Once again, these are closest to the counterclockwise values in the current investigation of 39.1, 26.8 and 30.6 Nmm but somewhat different than the values of 42.6, 27.2 and 43.3 Nmm for the same bracket systems in the clockwise direction. These results prove that both studies showed similar trends with the P-Dmn and A-Ovn brackets generating more torque than the A-Spd brackets. However, the clockwise results in the current study exhibited values around 10 Nmm higher for the P-Dmn and A-Ovn systems. This difference might be related to the direction of rotation performed in their study, which might have only been in a counterclockwise direction.

It is important to note the standard deviations (SD) listed in Table 2 with regards to each bracket system. The SD's for the P-Dmn and E-vic groups were consistently lower than the rest, even up at the maximum torquing angles in both directions. This could be due to their rigid bracket design and secure mechanisms of ligation. This also helps to confirm the repeatability of the methodology used in this study. However, looking at the other three systems presents a different scenario. All three of these active SLB systems present with larger SD's as high as 6.0 Nmm (with a coefficient of variation of 16%) for A-Emp at 30 degrees in the clockwise tests. This might indicate variability in the role of the active clip that provides assistance in wire engagement in the bracket-archwire complex, or possible deformation in the system at higher degrees of wire rotation. For this reason, active SLB systems might present with greater variation in torque expression from one system to the next.

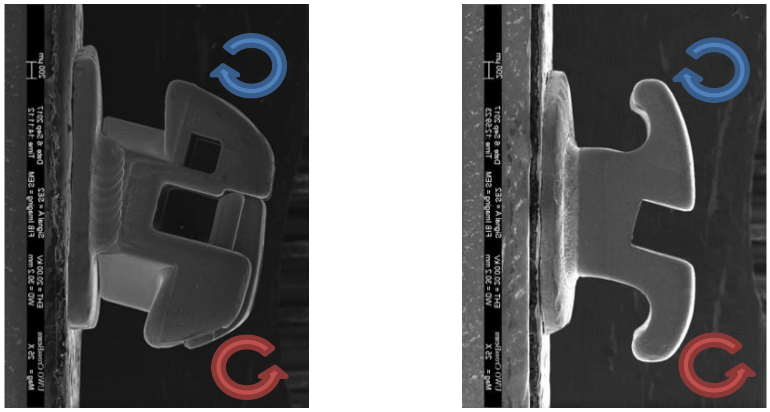
5.3 Direction of Wire Rotation

Unique to this study was the testing of all bracket systems in both the clockwise and counterclockwise directions of wire rotation. Most of the previous literature tested for torque in only one direction. Only two studies could be found that tested wire rotation in both directions.[5, 42] Unfortunately, these two studies focused mainly on the differences between 0.018-in and 0.022-in traditional twin bracket systems. Both studies used an OMSS to measure torque values and only reported measurements for -15 and +15 degrees of wire twist. For one of these studies, high-torque and low-torque brackets were bonded on the OMSS in an attempt to compare the differences, so the trends between direction of twist within a bracket system are hard to visualize.[5] For the other study, central incisor brackets with +22 degrees of torque prescription were used. The values at +15 and -15 degrees of wire twist were both 9.3 Nmm.[42] The twin bracket system in this current investigation generated values of 13.1 and 11.5 Nmm at the corresponding +15 and -15 degrees of wire twist, respectively. These values are relatively close with the slight differences likely explained by the different types of experimental set-ups. Using an OMSS allows for much more torque play or slop within the system. In addition, the long span of wire used in an OMSS is subject to less rigidity than that which is held tightly on either side by a clamping mechanism, such as the one used in the current investigation.

Comparing clockwise versus counterclockwise wire rotation within each individual bracket system reveals interesting insights. For the P-Dmn brackets, small but significant differences were found at low values probably due to the small SD. These differences would certainly not be considered clinically significant as most are under 2 Nmm. The differences in values above +24 degrees of wire rotation are under the 5 Nmm threshold so their clinical significance might also be questioned. Although these values are close, it is surprising that there was any difference at all, and this should be kept in mind when reviewing research studies of torque that do not report the direction of wire rotation.

The E-Vic brackets performed in a similar manner to the P-Dmn brackets. Although the values at +30 degrees of twist in the clockwise and counterclockwise tests are reaching almost three times that of maximum physiologic torque, it is not unreasonable to assume a clinician could place this much wire twist in a system. Therefore, the results for E-Vic at some of the higher degrees of wire twist could be both statistically and clinically significant as they are above 5 Nmm.

It was initially assumed that any passive SLB or traditional twin bracket system should have equal torquing capabilities in any direction due to the nature of the passive sliding door or elastomer. One explanation for the difference seen in this study may lie within the physical design of the bracket. Additional bulk of bracket material exists on the incisal portion of the bracket for P-Dmn and on the gingival portion for E-Vic (Figure 14 A and B). This could potentially lead to differences in the rigidity of the different parts of each bracket system, which might be more susceptible to deformation and energy loss in one direction of rotation more than the other. This potential for deformation might become more obvious when subjected to higher torquing forces such as those in this investigation.



(A) (B)

Figure 19: Scanning electron microscopy (SEM) images of a passive SLB and traditional twin bracket. Clockwise wire rotation is represented by blue arrows and counterclockwise rotation represented by the red arrows: A) SEM image of a P-Dmn bracket; B) SEM image of an E-Vic bracket.

The A-Ovn brackets recorded statistically significant differences at the smaller degrees of wire twist from -9 up until +9 degrees. It can be expected that these small differences under 3 Nmm would likely not prove clinically significant to most practitioners. At +24 degrees, a maximum difference of 12.7 Nmm was found and is likely clinically significant when considering the effective range of torque (5 – 20 Nmm). This is also an amount of twist (24 degrees) that is often applied in a clinical situation. Interestingly, this bracket system behaved the closest to the passive SLB and traditional twin bracket tested in the current study, as compared to the other active SLB's. This might indicate that the clip properties or design and rigidity of the bracket is similar to these systems.

When comparing different directions of rotation for the A-Spd bracket, the counterclockwise rotation generated higher torques with a difference just over 5 Nmm starting at +30 degrees, giving the potential for clinical significance. The difference continues to increase to a maximum difference of 19.6 Nmm at the maximum wire twist of +42 degrees. Additionally, the SD for A-Spd varied between directions of rotation.

During the clockwise rotations, the standard deviation rises steeply after approximately 30 degrees of wire twist. The behaviour of the counterclockwise rotation tests does not follow this pattern. A possible explanation for these changes in behaviour can be attributed to the active NT spring clip used with this bracket system. As the wire rotates in the clockwise tests, the active spring clip may play a more significant role than in the reverse wire twist direction. Due to the lack of stiffness with materials made of NT, this clip may start to deform or exert an inadequate force to support the wire at higher degrees of wire twist, leading to lower torque values and higher SD's in the clockwise direction of rotation. This phenomena has been observed in previous studies using this bracket system.[13, 19]

These observations with the Speed brackets were much more dramatic in a previous study conducted by Major et al[19]. It was discovered in their study that more than half of the Speed bracket doors opened during testing due to the high moments of torque. Although no doors were noted to open during testing for this investigation, it is possible that the design of the active NT clip played a large role in the torque differences observed between rotation directions. Anecdotally, it was noted that the doors for A-Spd (as well as A-Emp) were both more difficult to manipulate after undergoing a testing session. This could be related to significant door deformation but would require further imaging to verify.

The response to wire rotation direction for the A-Emp bracket was similar to what was observed in the A-Spd bracket, with significantly more torque generated in the counterclockwise direction starting at +12 degrees of rotation. The maximum difference of 16.7 Nmm noted at +42 degrees of rotation would have significant clinical implications. Again, these differences are likely a result of the active spring clip mechanism and design of the bracket system. Like the other active SLB system A-Spd, the spring clip seems to be more effected by clockwise rotations of the wire, resulting in lower overall torque in this direction of twist.

These unique results detailing the differences of wire twist direction in generating torque between some of the active SLB's have proven quite interesting. The results show that the Speed and Empower 2 Active systems are more capable of exerting lingual root torque as compared to buccal root torque, at higher degrees of archwire rotation. In addition to the asymmetric response of the clip mechanism, it is also speculated that these differences may arise from the design of the brackets with regards to the depth of the slot at the gingival wall. From the SEM photos in Figure 6, the differences in bracket slot gingival wall depths are readily evident between the passive SLB and twin bracket (Figure 6 A – B) versus the active SLB's (Figure 6 C – E), with A-Spd and A-Emp gingival walls being the shortest. Further investigation is necessary to determine the full effects of these differences on torque generation.

It is important to note that in a clinical situation, adjacent teeth would experience an average of the forces exerted as per Newton's third law. For example, applying buccal root torque to a central incisor would result in the adjacent teeth experiencing the same degree of lingual root torque due to equal and opposite forces within the entire bracket-archwire complex. Future studies evaluating torque generation should aim to report average values at each degree of rotation, or report values in both directions of wire rotation.

5.4 Clinically Relevant Torque Ranges

Clinically, the amount of torque that is desirable for biologically acceptable tooth movement has been reported to be between 5 and 20 Nmm. The P-Dmn bracket reached a value closest to the 5 Nmm threshold at 10 degrees, while reaching a value closest to the 20 Nmm threshold at 16 degrees, in the clockwise rotation tests (Table 4). This equates to a potential range of 6 degrees where the bracket would be subjecting a tooth to ideal physiologic tooth-moving forces. A-Ovn was similar, reaching these values at 10 and 16 degrees, as was E-Vic at 12 and 17 degrees. A-Spd reached 5 and 20 Nmm values a little later at 13 and 20 degrees whilst A-Emp required the greatest degree of wire rotation to reach this torque level at 15 and 23 degrees and also displayed the greatest range (8 degrees).

In the counterclockwise tests, the P-Dmn bracket reached a value closest to the clinically relevant 5 Nmm threshold again at 10 degrees of wire rotation, while reaching a value closest to 20 Nmm at 17 degrees (Table 5). This gives a potential range of 7 degrees for ideal torque force levels. E-Vic was similar, reaching 5 and 20 Nmm at 12 and 18 degrees, respectively. A-Ovn reached the thresholds at 13 and 20 degrees and A-Spd reached them at 14 and 21 degrees. A-Emp showed the greatest range in this direction as well, reaching the 5 and 20 Nmm values at 11 and 19 degrees (range of 8 degrees).

The results from both the clockwise and counterclockwise tests show a very small range of wire rotation applicable to healthy tooth movements in the 5 – 20 Nmm range (6 – 8 degrees). To the practicing clinician, this implies that care must be taken when measuring the amount of twist within the archwire before ligating the brackets to a conventional working wire. Additionally, the ranges may show more variation between bracket systems when studied using different wire compositions that would allow for a larger working range of rotations for 5 to 20 Nmm.

	5 Nmm Engagement (°)	20 Nmm Engagement (°)
P-Dmn	10	16
A-Spd	13	20
A-Ovn	10	16
A-Emp	15	23
E-Vic	12	17

Table 4: 5 Nmm and 20 Nmm engagement angles for the clockwise rotation tests.

	5 Nmm Engagement (°)	20 Nmm Engagement (°)
P-Dmn	10	17
A-Spd	14	21
A-Ovn	13	20
A-Emp	11	19
E-Vic	12	18

Table 5: 5 Nmm and 20 Nmm engagement angles for the counterclockwise rotation tests.

All three of the bracket systems in the Major et al.[19] study (P-Dmn, A-Spd and A-Ovn) expressed 20 Nmm of torque between 18 and 21 degrees of wire twist. In the current study, these same three bracket systems expressed 20 Nmm of torque between 16 and 20 degrees in the clockwise direction, and between 17 and 21 degrees in the counterclockwise direction. The results for the counterclockwise tests in this study show an almost identical range. The previous studies focused on reporting maximum torque values at high degrees of rotation and therefore had to extrapolate data for these ranges. It is possible that differences between bracket systems would be more noticeable if data was collected for every degree (or less), at these low angles of wire rotation.

It is important to note that there has been conflicting evidence in the literature to support the use of active over passive SLB systems, or vice versa, in generating torque for tooth movements. A recent systematic review showed minimal differences in terms of torque expression between active and passive SLB systems.[16] While some studies have found that the active brackets engage the archwires at an earlier stage of treatment (when smaller wires are used), they do not express much more torque than the passive brackets at higher degrees of twist in the wire.[13] Obviously, torque play is more exaggerated in the passive system compared to the active systems,[42] but studies have proven that this very minor amount of extra slop proves statistically insignificant at biologically sound levels of torque.[7, 13, 19, 54] Therefore, it has been proposed that the influence of the active or passive door mechanism is minimal and the size or precision of the slot dimensions are far more important for the transmission of torque to the tooth. It appears that the results of the present study support this statement, at least when using a conventional stainless steel working archwire.

Interestingly, it has been shown that there is no statistically significant difference between torque expression values in SLB systems when the doors are open or closed.[7] Although the amount of torque the spring clip exerts on the wire is only around 1 Nmm,[7] advocates of active self-ligation promote the idea that engagement of the bracket slot occurs sooner in the treatment sequence (with smaller dimension archwires), and thus for longer periods of time, rather than generating a higher torque value.[19] On the other hand, clinicians may argue that the little torque experienced in these early stages of the partially engaged wire range is insignificant to the final outcome of treatment.

5.5 Angles of Engagement and Torque Play

The angles of engagement and resulting torque play show differences between each bracket system. The results for P-Dmn show the least amount of play (15.5°) within the bracket-archwire system when the 0.019 x 0.025-in SS wire is engaged. This came unexpectedly, as passive SLB's are often thought to have the greatest amount of slop due to their method of ligation. It can only be speculated that the reason for these results may lie within the rigidity of the sliding door, and/or because of an accurate manufacturing process leading to a more intimate relationship between the slot and archwire. The torque play calculated for A-Ovn and E-Vic were similar at 18° and 18.5°, respectively. These results suggest that once a conventional SS working wire is used, a passive SLB system (such as Damon Q) can have less torque play and be able to generate more torque than a traditional twin or active SLB system.

The torque play for A-Spd and A-Emp were the highest at 20.5° and 21.5°, respectively. This came as a surprise as it was hypothesized that these two systems would have the least amount of slop because of their active clip mechanisms. It is speculated that the slop in these active SLB's may be a result of the clips being unable to exert forces capable of displacing the wire into the bracket slot base with such a large, rigid wire. A different outcome may have been realized using smaller wires with a lower modulus of elasticity, where the active clip may have been more effective in facilitating torque generation.

5.6 Strengths of this Study

Many strengths of this study are readily observed. The first strength lies in the diversity of the bracket systems tested. To date, there has not been any published literature testing more than three SLB systems within the same study. Additionally, there has yet to be any literature comparing all three types of ligation in the same study (traditional twin brackets, passive SLB's and active SLB's). The fact that this study includes all three bracket types and five total bracket systems allows for diverse comparisons between some of the more common contemporary bracket systems used today.

Another strength of the study was in testing both directions of wire rotation (clockwise and counterclockwise). Very few studies have examined the effect of wire twist direction, let alone while using different types of bracket systems. Although there is little data to be found in the literature to compare the results, they suggest that future studies looking at torque should report both directions of wire rotation and potentially report an average of both directions as well.

The greatest strength of this study likely lies in the methodology, along with the fabrication and implementation of the custom set-up. The mounting jig and custom torque assembly are unique to the present study and were meticulously crafted to carefully position the brackets and wires in the same location for each test while maintaining absolute rigidity. In addition, these devices were fabricated to be used with a new model of an Instron machine (Instron Electropuls E10000; Norwood, MA, USA). The company was able to provide independent testing of the accuracy of the current set-up, which previous studies have not reported. It is well known that this company and its products are of very high quality and used worldwide for accurate material testing in various fields of research. This novel approach to testing torque generation has proven to provide accurate results and is easily adaptable to future studies using different brackets or wires.

5.7 Limitations of this Study

A few limitations of this study become evident when closely examining the methodology. The first limitation involves the selection of the 10 kN load cell used in this investigation which is ideally most suited for experiments measuring much higher moments. As explained in the methodology, the amount of jitter or noise that could potentially be realized at the target value of 20 Nmm brings to question the accuracy of some of the results, particularly at the lower levels of torque used in the clinically relevant range. However, the average error value reported through independent lab testing at the Instron facility (2.86 Nmm) was adequate for this study and capable of detecting small differences in torque values between groups.

Another limitation exists within the manipulation of the Instron machine itself. Although centering of the wire and bracket slot was done by careful visual manipulation, a small level of operator error cannot be ruled out. In addition, there was potential for minute alterations in starting points due to hysteresis in the system. To address these potential problems, an attempt was made to neutralize the effect of any uneven centering of the archwire by measuring in both the positive and negative wire rotation directions for each test and adjusting the data. It would have been ideal to find a way to perfectly center the bracket-archwire complex within the set-up itself for each test, instead of addressing this issue after data collection.

The methodology in this study used to determine the angles of engagement for each bracket system was subjective. Manual extrapolation through visualization was performed in an effort to note the areas on the graphs where the slope changed considerably. Using this method, torque play could only be based on whole degree values at locations that seemed most reasonable for a magnitude in change indicating bracket-archwire engagement. These values must be interpreted as rough evaluations of the actual system and what could potentially be expected. Ideally, a software program should be developed that can precisely measure the slope at every degree in order to accurately determine a point of significant slope change.

The traditional twin bracket system was ligated with a new, unstretched and dry elastomer. In this state, the wire is ligated securely to the bracket but does not simulate an intraoral environment. At higher temperatures and pH levels, the elastomers will degrade causing a loss of secure ligation and potentially different torque generation.[31] Over time, the torque expression may change drastically under these conditions.[31, 42, 47] This concept may also potentially apply to the active clip mechanisms, depending on their material compositions. It would have been ideal to perform these tests in an environment replicating the mouth, over a longer period of time, in order to further understand the effects of different ligation methods.

A final critique of this study might be made about the fact that the sample size per group was lower than some previous studies in the literature. However, the power of the present study was more than adequate to detect significant differences between groups at +42, +24 and +12 degrees of wire rotation (Power=1.000). As such, this sample size was adequate, and the results will further serve as data for future sample size calculations in relation to any follow-up studies.

5.8 Suggestions for Future Research

Future research could involve different wire sizes and compositions. Although this study proved insightful to conventional SS working wires in an 0.022-in bracket slot, there are additional wire sizes that are used frequently and that might exert different torque on teeth. Testing conventional TMA working wires, along with smaller NT, TMA and SS wires would prove beneficial to the clinician when choosing the proper archwire sequence for each patient. Part of the claims made by active SLB manufacturers includes the expression of torque during the initial stages of treatment when smaller rectangular wires are used. It may prove true that the active clip mechanisms in these systems are able to generate physiologic force levels at lower levels of wire rotation in these smaller wires. Until tests are done with smaller wire sizes, it is unknown if these claims of providing better torque expression with smaller wires is true.

Another future study could focus on bracket deformation at different degrees and directions of wire rotation. This could prove helpful to confirm some of the current investigation's speculations of torque expression variation based on bracket design. It might also help explain the results from this study showing differences in the passive self-ligation (P-Dmn) and twin bracket (E-Vic) systems when rotated in different directions.

Finally, testing even more bracket systems and types can only provide the clinician with additional information when choosing their fixed appliances. Many other contemporary SLB and traditional twin bracket systems are used frequently in North America and would prove interesting to compare side-by-side with those already tested. In addition, some bracket systems claim to come in unique slot sizes that exert more torque at smaller angles of twist. Obviously, to test these systems against those already known would help to identify the truthfulness of such claims by the manufacturer.

Chapter 6

6 Conclusions

An *in vitro* study was conducted to compare the torque values generated by a conventional stainless steel working archwire engaged in three contemporary active SLB bracket systems, a contemporary passive SLB bracket system and a traditional twin bracket system with elastic ligatures. It was found that:

1. The twin bracket (E-Vic), passive SLB (P-Dmn), and one of the active SLB's (A-Ovn) generated similar torque values, significantly higher than the other two active SLB's (A-Spd and A-Emp) at most degrees of wire rotation.
2. Two of the active SLB's (A-Spd and A-Emp) generated significantly lower torque in the clockwise versus the counterclockwise direction of rotation at higher degrees of wire rotation.
3. All five bracket systems had similar ranges of clinically relevant torque (5 – 20 Nmm), ranging from 16 to 23 degrees of rotation.

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Appendices

Appendix A: Complete set of unadjusted mean torque values (Nmm) from -15 to +45 degrees of rotation.

	Torque (Nmm)									
	Damon		Speed		InOvation-R		Empower		Victory	
Angle(°)	Clock	Counter	Clock	Counter	Clock	Counter	Clock	Counter	Clock	Counter
-15	5.8	16.6	1.5	8.2	9.7	16.2	11.5	7.2	11.1	13.0
-14	3.8	14.3	1.2	6.1	7.3	13.6	9.3	6.0	8.7	10.0
-13	2.0	11.4	0.9	4.7	5.1	10.9	7.1	4.6	6.3	7.4
-12	1.0	9.1	0.8	3.2	3.4	8.3	5.4	3.6	4.6	5.4
-11	0.7	7.0	0.6	2.3	2.1	6.3	3.9	3.1	3.0	3.8
-10	0.7	5.0	0.5	1.9	1.4	4.7	3.1	2.7	2.2	3.1
-9	0.5	3.5	0.6	1.5	1.2	3.8	3.0	2.3	1.8	2.5
-8	0.4	2.3	0.4	1.3	1.2	3.1	3.0	2.5	1.7	2.2
-7	0.3	1.7	0.4	1.1	1.0	2.9	2.8	2.4	1.6	2.1
-6	0.3	1.5	0.3	0.9	0.9	2.6	2.6	2.5	1.6	1.8
-5	0.6	1.3	0.3	0.7	0.9	2.6	2.3	2.4	1.5	1.8
-4	0.7	1.2	0.3	0.8	0.9	2.4	1.9	2.5	1.2	1.8
-3	0.8	1.0	0.4	0.7	0.9	2.2	1.5	2.5	1.3	1.4
-2	0.8	0.9	0.2	0.8	0.7	2.0	1.1	2.5	1.1	1.4
-1	0.9	0.6	0.3	0.6	0.4	1.7	0.8	2.1	0.9	1.1
0	1.1	0.5	0.4	0.4	0.7	0.8	0.8	1.2	0.5	0.7
1	2.0	0.2	0.9	0.3	1.7	0.5	1.7	1.4	1.7	1.3
2	2.2	0.3	1.0	0.4	2.0	0.8	2.1	1.8	1.9	1.5
3	2.5	0.4	1.0	0.6	2.2	0.9	2.6	2.1	1.9	1.3
4	3.5	0.4	1.1	0.6	2.0	1.0	2.5	2.3	2.1	1.5
5	4.5	0.6	1.3	0.7	2.0	1.2	2.9	2.3	2.1	1.5
6	6.5	0.7	1.6	0.7	2.2	1.2	3.0	2.4	2.3	1.6
7	8.6	1.1	2.3	0.9	2.5	1.3	2.9	2.7	2.2	1.5
8	11.0	1.8	3.7	1.0	2.7	1.2	2.5	2.8	2.2	1.9
9	13. 6	3.2	5.3	1.0	3.4	1.3	2.7	2.9	2.3	2.1
10	16.1	5.1	7.3	1.2	4.8	1.4	2.6	3.1	2.6	2.8
11	18.8	7.3	9.3	1.7	6.8	2.3	2.7	3.5	3.5	3.9
12	21.6	9.2	11.5	2.6	9.0	3.5	2.9	4.2	5.3	5.4
13	24.4	11.6	13.6	3.9	11.2	5.1	3.4	4.8	7.6	7.4
14	27.3	13.9	15.5	5.4	14.1	6.9	4.1	5.9	10.1	9.4
15	30.3	16.4	17.8	6.9	16.9	9.1	5.2	7.5	13.1	11.5
16	33.1	18.6	19.7	8.7	19.7	11.4	6.4	9.1	16.1	14.0
17	36.4	20.9	21.8	10.7	22.7	13.6	8.1	11.2	19.2	16.4
18	39.4	23.3	23.7	12.9	25.5	16.1	9.8	13.2	22.4	18.8
19	42.6	25.9	25.4	15.1	28.4	18.5	11.9	15.4	25.5	21.2
20	45.6	28.1	27.2	17.5	31.5	20.8	14.0	17.6	28.8	23.5
21	48.7	30.7	28.7	19.8	34.5	23.3	16.3	19.8	32.2	25.9

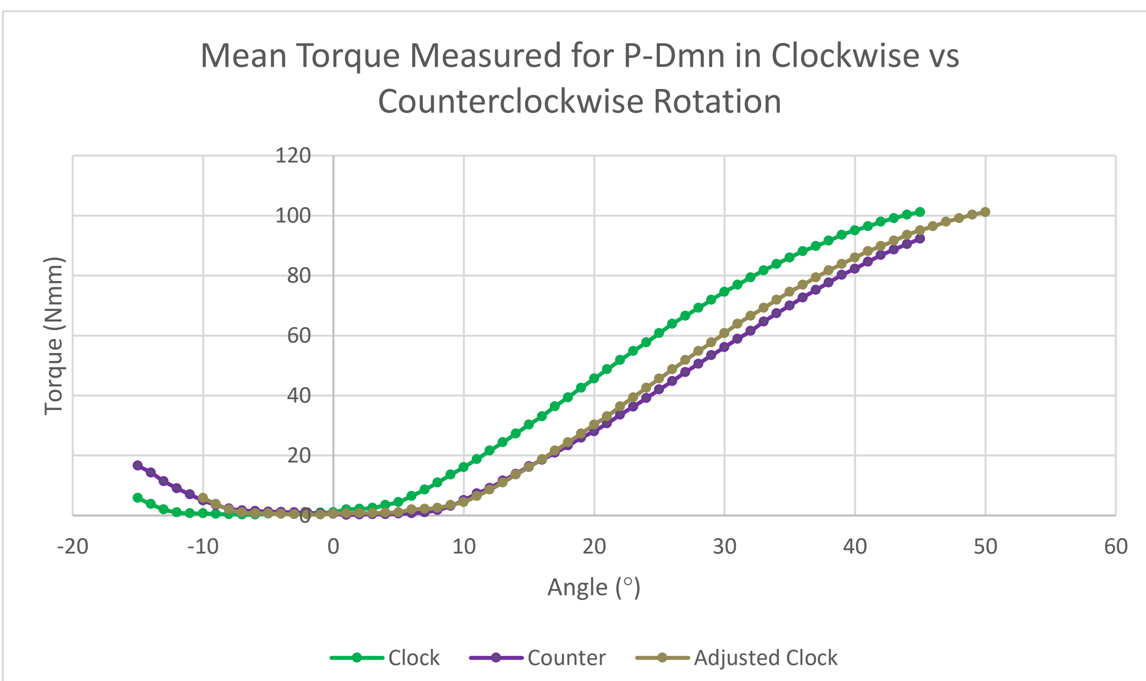
22	51.8	33.6	30.6	22.0	37.3	25.7	18.5	22.1	35.6	28.3
23	54.8	36.3	32.4	24.4	40.3	28.1	20.8	24.3	39.0	30.8
24	57.7	39.1	34.1	26.8	43.3	30.6	23.2	26.5	42.5	33.3
25	60.8	42.0	35.8	29.2	46.3	33.3	25.5	28.5	45.7	36.2
26	63.9	44.8	37.5	31.9	49.2	36.4	27.9	31.1	49.1	39.2
27	66.6	47.8	39.2	34.9	52.0	39.4	30.3	33.5	52.3	42.2
28	69.2	50.6	41.0	37.6	54.9	42.4	32.6	36.1	55.4	45.2
29	71.9	53.4	42.8	40.4	57.7	45.5	35.2	39.0	58.6	48.2
30	74.6	56.1	44.6	43.4	60.5	48.4	37.4	41.7	61.6	51.1
31	76.9	58.9	46.5	46.2	63.2	51.3	39.8	44.7	64.8	54.1
32	79.4	61.5	48.1	49.1	65.8	54.3	42.3	47.7	67.6	56.8
33	81.7	64.6	49.9	52.1	68.2	57.1	44.5	50.6	70.4	60.0
34	83.9	67.4	51.5	55.0	70.7	60.5	46.9	53.7	72.9	63.0
35	86.0	70.0	52.9	57.9	73.1	63.6	49.1	56.6	75.7	65.9
36	88.1	72.7	54.4	60.7	75.5	66.3	51.3	59.7	78.3	68.7
37	89.8	75.2	55.8	63.6	77.9	69.3	53.6	62.5	80.7	71.4
38	91.6	77.7	57.2	66.3	80.2	72.0	55.6	65.6	83.2	74.3
39	93.6	80.2	58.4	69.1	82.0	74.8	57.9	68.2	85.4	76.9
40	95.1	82.3	59.3	71.7	84.2	77.4	59.6	70.7	87.5	79.4
41	96.4	84.6	59.9	74.4	86.2	79.9	61.9	73.3	89.3	81.7
42	97.9	86.8	60.4	76.8	87.9	82.3	63.9	76.0	91.2	84.1
43	99.1	88.7	61.0	79.3	89.8	84.6	65.9	78.5	93.0	86.4
44	100.3	90.5	61.6	81.4	91.4	86.7	67.8	80.6	94.8	88.5
45	101.1	92.3	61.9	83.5	93.0	88.8	67.6	82.9	95.0	90.3

Appendix B: Complete set of adjusted mean torque values (Nmm) from -9 to +42 degrees of rotation.

	Torque (Nmm)									
	Damon		Speed		InOvation-R		Empower		Victory	
Angle(°)	Clock	Counter	Clock	Counter	Clock	Counter	Clock	Counter	Clock	Counter
-9	3.8	3.5	0.9	1.5	1.2	3.8	3.0	2.4	1.8	2.5
-8	2.0	2.3	0.8	1.3	1.2	3.1	3.0	2.5	1.7	2.2
-7	1.0	1.7	0.6	1.1	1.0	2.9	2.8	2.4	1.6	2.1
-6	0.7	1.5	0.5	0.9	0.9	2.6	2.6	2.5	1.6	1.8
-5	0.7	1.3	0.6	0.7	0.9	2.6	2.3	2.5	1.5	1.8
-4	0.5	1.2	0.4	0.8	0.9	2.4	1.9	2.5	1.2	1.8
-3	0.4	1.0	0.4	0.7	0.9	2.2	1.5	2.1	1.3	1.4
-2	0.3	0.9	0.3	0.8	0.7	2.0	1.1	1.2	1.1	1.4
-1	0.3	0.6	0.3	0.6	0.4	1.7	0.8	1.4	0.9	1.1
0	0.6	0.5	0.3	0.4	0.7	0.8	0.8	1.8	0.5	0.7
1	0.7	0.2	0.4	0.3	1.7	0.5	1.7	2.1	1.7	1.3
2	0.8	0.3	0.2	0.4	2.0	0.8	2.1	2.3	1.9	1.5
3	0.8	0.4	0.3	0.6	2.2	0.9	2.6	2.3	1.9	1.3
4	0.9	0.4	0.4	0.6	2.0	1.0	2.5	2.4	2.1	1.5
5	1.1	0.6	0.9	0.7	2.0	1.2	2.9	2.7	2.1	1.5
6	2.0	0.7	1.0	0.7	2.2	1.2	3.0	2.8	2.3	1.6
7	2.2	1.1	1.0	0.9	2.5	1.3	2.9	2.9	2.2	1.5
8	2.5	1.8	1.1	1.0	2.7	1.2	2.5	3.1	2.2	1.9
9	3.5	3.2	1.3	1.0	3.4	1.3	2.7	3.5	2.3	2.1
10	4.5	5.1	1.6	1.2	4.8	1.4	2.6	4.2	2.6	2.8
11	6.5	7.3	2.3	1.7	6.8	2.3	2.7	4.8	3.5	3.9
12	8.6	9.2	3.7	2.6	9.0	3.5	2.9	5.9	5.3	5.4
13	11.0	11.6	5.3	3.9	11.2	5.1	3.4	7.5	7.6	7.4
14	13. 6	13.9	7.3	5.4	14.1	6.9	4.1	9.1	10.1	9.4
15	16.1	16.4	9.3	6.9	16.9	9.1	5.2	11.2	13.1	11.5
16	18.8	18.6	11.5	8.7	19.7	11.4	6.4	13.2	16.1	14.0
17	21.6	20.9	13.6	10.7	22.7	13.6	8.1	15.4	19.2	16.4
18	24.4	23.3	15.5	12.9	25.5	16.1	9.8	17.6	22.4	18.8
19	27.3	25.9	17.8	15.1	28.4	18.5	11.9	19.8	25.5	21.2
20	30.3	28.1	19.7	17.5	31.5	20.8	14.0	22.1	28.8	23.5
21	33.1	30.7	21.8	19.8	34.5	23.3	16.3	24.3	32.2	25.9
22	36.4	33.6	23.7	22.0	37.3	25.7	18.5	26.5	35.6	28.3
23	39.4	36.3	25.4	24.4	40.3	28.1	20.8	28.5	39.0	30.8
24	42.6	39.1	27.2	26.8	43.3	30.6	23.2	31.1	42.5	33.3
25	45.6	42.0	28.7	29.2	46.3	33.3	25.5	33.5	45.7	36.2
26	48.7	44.8	30.6	31.9	49.2	36.4	27.9	36.1	49.1	39.2
27	51.8	47.8	32.4	34.9	52.0	39.4	30.3	39.0	52.3	42.2
28	54.8	50.6	34.1	37.6	54.9	42.4	32.6	41.7	55.4	45.2
29	57.7	53.4	35.8	40.4	57.7	45.5	35.2	44.7	58.6	48.2
30	60.8	56.1	37.5	43.4	60.5	48.4	37.4	47.7	61.6	51.1

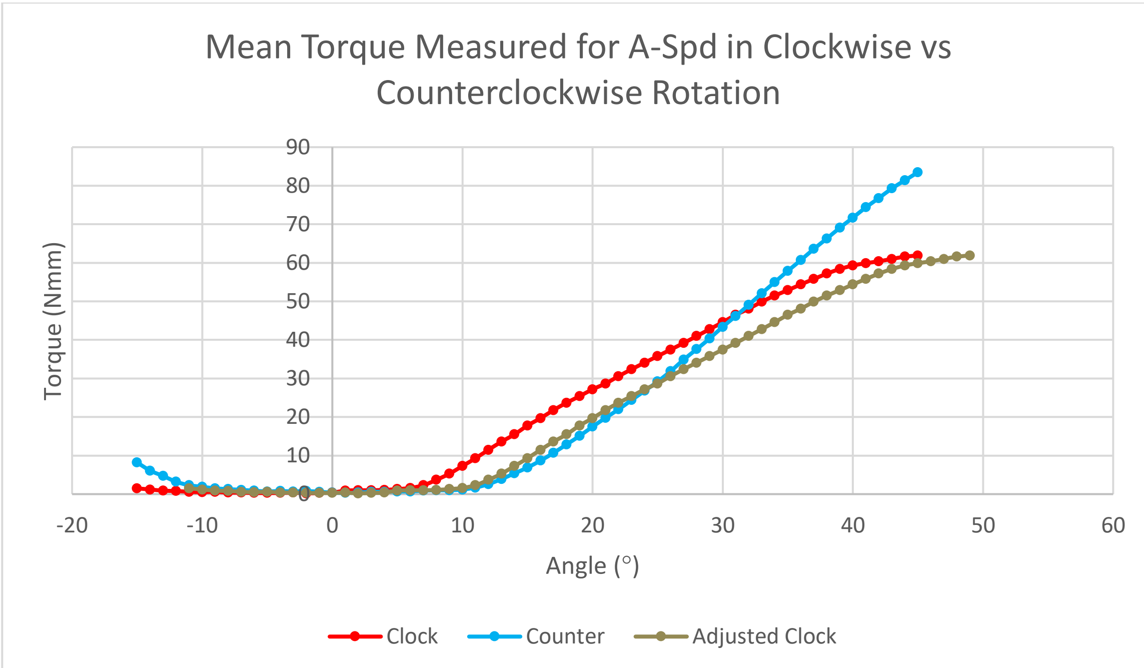
31	63.9	58.9	39.2	46.2	63.2	51.3	39.8	50.6	64.8	54.1
32	66.6	61.5	41.0	49.1	65.8	54.3	42.3	53.7	67.6	56.8
33	69.2	64.6	42.8	52.1	68.2	57.1	44.5	56.6	70.4	60.0
34	71.9	67.4	44.6	55.0	70.7	60.5	46.9	59.7	72.9	63.0
35	74.6	70.0	46.5	57.9	73.1	63.6	49.1	62.5	75.7	65.9
36	76.9	72.7	48.1	60.7	75.5	66.3	51.3	65.6	78.3	68.7
37	79.4	75.2	49.9	63.6	77.9	69.3	53.6	68.2	80.7	71.4
38	81.7	77.7	51.5	66.3	80.2	72.0	55.6	70.7	83.2	74.3
39	83.9	80.2	52.9	69.1	82.0	74.8	57.9	73.3	85.4	76.9
40	86.0	82.3	54.4	71.7	84.2	77.4	59.6	76.0	87.5	79.4
41	88.1	84.6	55.8	74.4	86.2	79.9	61.9	78.5	89.3	81.7
42	89.8	86.8	57.2	76.8	87.9	82.3	63.9	80.6	91.2	84.1

Appendix C: Unadjusted and Adjusted mean P-Dmn torque values.



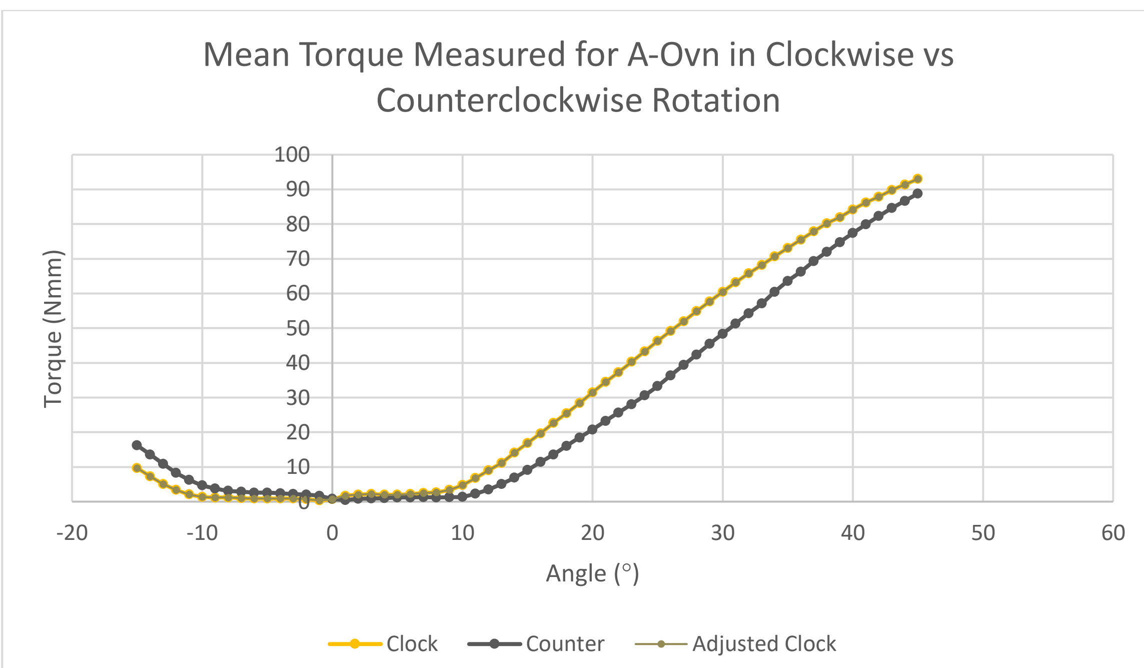
**P-Dmn Clock values adjusted by adding 5 degrees at each measurement*

Appendix D: Unadjusted and Adjusted mean A-Spd torque values.



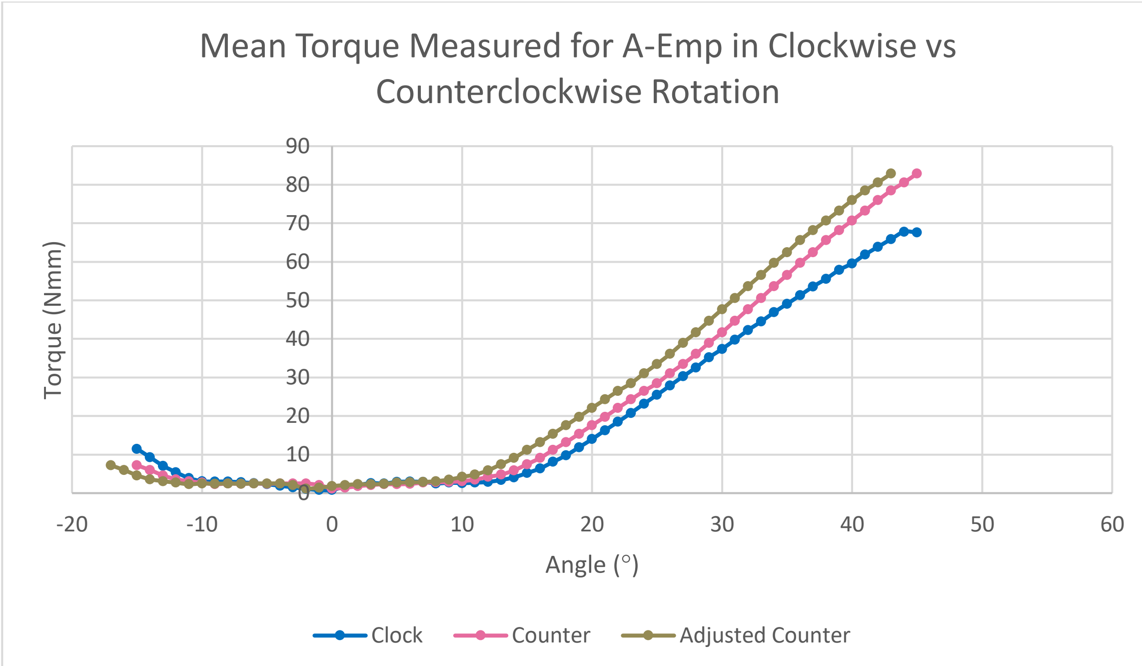
**A-Spd Clock values adjusted by adding 4 degrees at each measurement*

Appendix E: Unadjusted and Adjusted mean A-Ovn torque values.



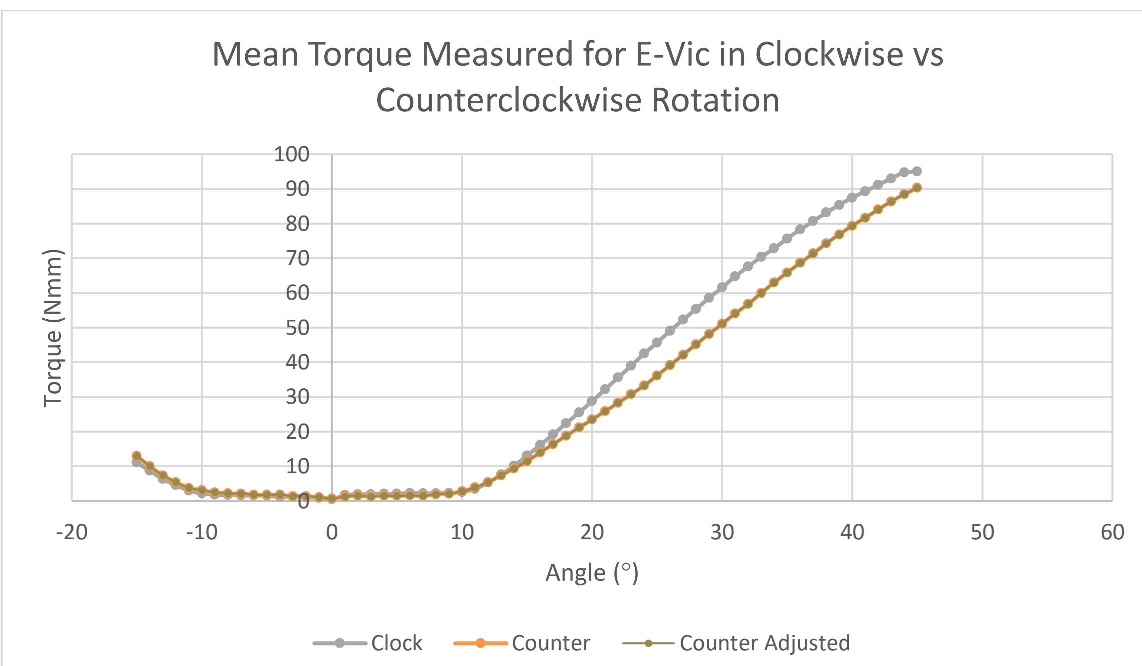
**A-Ovn did not need adjustment of original measurements*

Appendix F: Unadjusted and Adjusted mean A-Emp torque values.



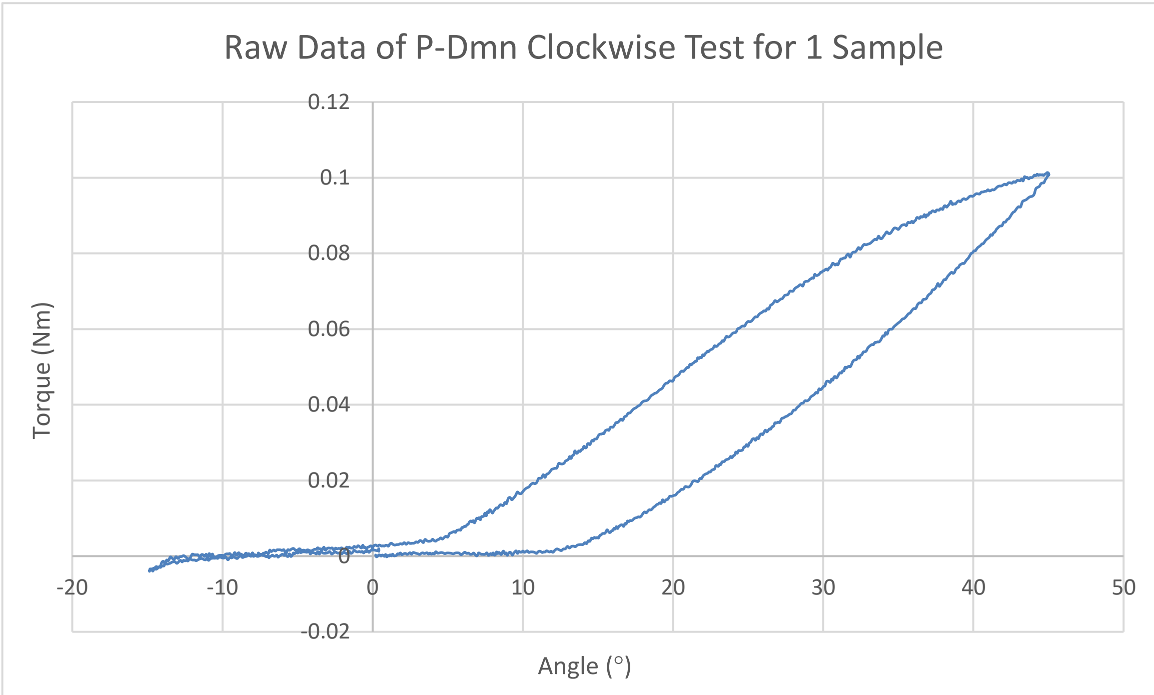
**A-Emp Counter adjusted by subtracting 2 degrees at each measurement*

Appendix G: Unadjusted and Adjusted mean E-Vic torque values.



**E-Vic did not need adjustment of original measurements*

Appendix H: Raw data of a single P-Dmn sample tested in the clockwise direction, showing hysteresis as the difference in starting and ending points at zero degrees.



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