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First Order Archwire Deflections: Comparing Conventional, Active Self-Ligating and Passive Self-Ligating Mechanisms

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First order archwire deflections.
Comparing conventional, active self-ligating and passive self-ligating mechanisms.

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First order archwire deflections. Comparing conventional, active self-ligating and passive self-ligating brackets

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Abstract

Introduction: Our aim in this study was to compare conventional, active self-ligating and passive self-ligating mechanisms using first order archwire deflections. A cantilever and three point bend model were used to assess degree of bracket play, deactivation force, and superelastic range. **Methods:** We studied five different bracket types: *Victory*^{TM} *(* 3M-Unitek, St. Paul, MN), *SmartClip*^{TM} (3M-Unitek, St. Paul, MN), *Damon MX*^{TM} (Ormco, Glendora, CA), *Carriere*^{TM} (Ortho Organizers, Carlsbad, CA) and *In-Ovation R*^{TM} (GAC Intl., Bohemia, NY). These brackets were tested with round superelastic nickel titanium wires varying in diameter: 0.014", 0.016", 0.018" and 0.020" (Ultimate Wireforms Inc., Bristol, CT). Using a mechanical testing apparatus with custom fixture we recorded the load-deflection curve generated by the deflected archwire over a span of 4mm. The wire was deflected at a rate of 10mm/min and measurements were taken at a rate of 20Hz. Data was compared and analyzed using a single factor ANOVA with the bracket type as the discriminating variable. Group differences were further analyzed using post-hoc T-tests ($\alpha = 0.01$). **Results:** Bracket play was significant for both wire diameters ($p \leq 0.01$). The 0.014" wire showed significantly less play with the Victory and In-Ovation R brackets when compared to the passive self ligating brackets (PSLBs). For PSLBs, the SmartClip bracket had significantly less bracket play than the Damon MX and Carriere, which were not significantly different. Both the ligation mechanism and the archwire diameter had a significant effect on deactivation load. Victory and In-Ovation R brackets provided a measurable deactivation load at 1mm, while the PSLBs provided no such force with the 0.014" wire. During the cantilever test, PSLBs produced the lowest deactivation forces for all wires tested. The

three point bend test showed similar results with small diameter wires. As the degree of deflection and archwire diameter increased, the PSLBs displayed reduced deactivation forces, which are related to significant binding and increased sliding frictional resistance. The superelastic range varied significantly for the various bracket types ($p \leq 0.01$). Victory provided the largest superelastic range for all wires except the 0.020". Of the SLBs tested, In-Ovation R displayed the larger superelastic range. The 0.014" test showed a significant difference in superelastic range for all brackets except the Damon MX and Carriere, which were not significantly different from each other. **Conclusions:** The type of ligation has a significant effect on the degree of bracket play, force of deactivation and superelastic range. These factors influence the clinician's ability to achieve full first-order correction and to maintain light, continuous forces over a wide range of activation. The clinician should consider the degree of correction needed and type of ligation when selecting the appropriate alignment wire.

Introduction

We have observed clinically that a rotated tooth, when ligated to a straight archwire, does not rotate about its center of resistance, but instead rotates about an axis located at or near the bracket of the tooth in motion (See Figure 1). If only a pure moment was present at the bracket/archwire interface we would expect rotation about the center of resistance, but this is not the case. The type of movement observed, translation of the center of resistance and rotation about the bracket, requires the presence of both a moment and a force. In the case of a rotated tooth, the activation of the archwire into the bracket slot provides a buccolingual couple responsible for the rotational moment (See Figure 1). But how can we explain translation of the center of resistance? We surmise that the force responsible for translating the center of resistance in a mesiodistal direction is the force of friction acting opposite the direction of derotation. This leads us to question the role of ligation in first order orthodontic tooth movement and whether variations in bracket ligation type affect the first order force system. We plan to test this hypothesis by assessing the load-deflection curves for conventional, active self-ligating and passive self-ligating brackets when subject to first order archwire deflections.

Background

Orthodontic Tooth Movement

Orthodontic tooth movement is accomplished through the application of force to the supportive structures of the teeth. The fixed orthodontic appliance, which is responsible

for the delivery of force, is selected and activated by the clinician. The clinician determines the magnitude and direction of force to be applied to the dentition. Activation of the orthodontic appliance produces stress and strain within the supportive structures of the teeth. These supportive structures are insensitive to the bracket design, wire shape or alloy being used in the orthodontic appliance. Instead, their response is based solely on the stresses and strains occurring in the local environment. To achieve a precise biologic response, precise orthodontic stimuli must be applied. Even though there is variability among individuals' biologic response, understanding biomechanics and the delivery of orthodontic force improves treatment by reducing variability in the expected treatment outcome [1].

Anatomy and the Biological Response:

The supportive structures of the teeth consist of the alveolar bone, periodontal ligament (PDL), mesenchymal cells and ground substance. The alveolar bone surrounds the tooth root and is covered by a thin layer of cortical bone called the lamina dura. From the lamina dura extend the collagenous fibers of the periodontal ligament. These fibers are embedded in alveolar bundle bone on one side, extend across the 0.5mm ligament space and attach to the cementum layer of the tooth root on the other side. Additionally, the PDL space contains a network of capillaries and nerve fibers, as well as an amorphous ground substance consisting of connective tissue polysaccharides, salts and water.

The orientation of the collagenous fiber bundles of the PDL varies with the functional demands of the dentition. The majority of fibers, the oblique fibers, run apically from the

alveolar bone to the cementum (See Figure 2). This arrangement of fibers functions as a "shock absorber", enabling teeth to withstand the forces of normal function. When forces are applied to the teeth, the underlying PDL fibers, cells, interstitial fluid and alveolar bone flex to dissipate the stress.

Though the alveolar bone is constantly remodeling in response to functional demands, the high magnitude and short duration of functional forces is inadequate to produce tooth movement. Studies by van Leeuwen et al. found that continuous forces are more conducive to orthodontic tooth movement than discontinuous forces [2](See Figure 3). Light, continuous compression and tension within the PDL space stimulates a sequence of events that initiates remodeling of the surrounding alveolar bone. These stresses alter the local fluid pressure and vary the blood flow in the PDL. This change in pressure and blood flow leads to the release of chemical mediators. These chemical mediators initiate a cascade of signals that lead to the activation of osteoclasts and osteoblasts, the primary bone remodeling cells.

The remodeling process consists of concomitant resorption and apposition of the alveolar bone surrounding the tooth root. Bone resorption is the result of increased osteoclast activity and occurs in areas of compression of the PDL space. Bone deposition occurs in areas of tension and is a result of increased osteoblast activity [3]. The osteoclasts create space in the alveolar bone for the tooth to move, while the osteoblasts form new bone in the areas vacated by the moving tooth. This process occurs in three dimensions about the

tooth root and is a product of the local and systemic biological factors, as well as the magnitude, duration and direction of the externally applied force.

Biomechanics of Tooth Movement

The orthodontic appliance is designed to move teeth by applying a force system at a distance from the center of resistance. For a free body, the center of resistance is equivalent to the center of mass (See Figure 4). When a force is applied through the center of resistance a free body will move linearly without rotation. Teeth, though, are embedded in alveolar bone and restrained by the periodontal ligament. Hence, the center of resistance is not analogous with the center of mass. In a healthy periodontium, the center of resistance of a tooth is located below the crest of the alveolar bone. Proffit states that its location can be estimated by taking a point halfway between the midpoint of the embedded root and the crest of the alveolar bone [4]. Burstone calculates the location of the center of resistance of a single rooted tooth by multiplying the distance between the root apex and alveolar crest by 0.33 [5]. As evident in both of these estimates, the center of resistance is dependent on the root length, morphology and the level of the alveolar bone (See Figure 5).

Any force applied at a distance from the center of resistance will produce a moment of the force (M_F)(See Figure 6) [6]. The moment of the force (M_F) can be calculated by multiplying the force magnitude by the perpendicular distance from the center of resistance, M_F = Force x distance. This force system will produce simultaneous linear and rotational movement. Variation of either the force magnitude or the distance from the center of resistance can have a significant affect on the magnitude of M_F.

Due to the center of resistance of the tooth being at a distance from the point of force application, the orthodontic appliance must use couples to control tooth movement in three dimensions. A couple, or pure moment, will produce pure rotation about the center of resistance. A couple is defined as two parallel forces of equal magnitude acting in opposite directions and separated by a distance [1]. Regardless of where the couple is applied it will produce rotation about the center of resistance (See Figure 7). Couples are often referred to as the applied moment. In the case of fixed orthodontic appliances, these couples occur between the bracket and archwire and/or the ligature and the archwire. As tooth movement occurs, couples serve to counteract moments of the applied forces. The moment of the force will tend to rotate the crown in the direction of force application, but couples within the bracket system function to offset the magnitude of the moment of the force. By offsetting the moment of the force, these couples aid in maintaining the orientation of the tooth as the center of resistance moves linearly.

We can predict tooth movement by combining the force, moment of the force, and applied moment into an equivalent force system. This equivalent force system defines a force system at the center of resistance that is equivalent to the force system being applied at the bracket. As seen in Figure 8, the sum of the forces and moments are replaced at the center of resistance. The net force is the vector sum of forces being applied at the bracket, while the net moment is determined by adding the moment of the force and the applied moment of the couple. The ratio of this net moment to the net force (M/F) can be used to predict tooth movement and the center of rotation of the tooth [6-8].

There are four basic types of tooth movement: tipping, translation, root movement and rotation. Each type of movement can be defined by its specific moment/force ratio. Tipping occurs when the center of rotation is apical to the center of resistance. There are two types of tipping: controlled and uncontrolled (See Figure 9). Uncontrolled tipping occurs when the ratio of the $M/F \leq 5:1$, which places the center of rotation between the center of resistance and the root apex. Controlled tipping occurs when the ratio of the $M/F = 7:1$, which places the center of rotation at the root apex [6]. Controlled tipping requires a couple to control for the moment of the force, while uncontrolled tipping can occur without a compensating couple.

Translation, also known as "bodily movement", occurs when the center of rotation is located at an infinite distance from the center of resistance (See Figure 10). Typically this type of movement requires an $M/F = 10:1$ [6]. Translation can be achieved in two ways. One is by applying a force through the center of resistance of the tooth, which can be difficult because it is usually located below the crest of alveolar bone. The other is to incorporate an applied moment that negates the moment of the force. In this situation there is a net linear force through the center of resistance but no net rotation.

The third type of tooth movement is root movement. This occurs when the center of rotation is located coronal to the center of resistance. Root movement requires an $M/F \geq 12:1$ [6]. With this type of movement the root apex of the tooth will move a greater distance than the crown (See Figure 10). Clinicians frequently refer to this root movement as "torque".

The final type of movement is rotation. Pure rotation, which requires the absence of linear movement of the center of resistance, occurs when the center of rotation is located at the center of resistance. Pure rotation requires the presence of a couple. It cannot be achieved through the combination of a moment and a force. Without an equal and opposite force there will always be a linear effect on the center of resistance, therefore producing translation in combination with the desired rotational movement (See Figure 11).

Fundamentals of Static Equilibrium

Static equilibrium implies that the sum of the forces and moments acting on a body is zero. This concept is derived from Newton's three laws of motion. The first law, called the Law of Inertia, states that a body at rest will remain at rest and a body in motion will remain in motion unless acted upon by an external force. This means that a body at rest is at rest because the sum of all forces acting on the body equals zero. The second law, Law of Acceleration, deems that force is equal to mass times acceleration (F=ma) and that a force will act to change motion in the direction that it acts. Newton's third law of motion is the Law of Action and Reaction. This law states that to every action there is an equal but opposite reaction. This third law is fundamental to orthodontic biomechanics because it stresses that any force applied to achieve a desired outcome will have a reactive force capable of an undesirable outcome.

When evaluating an orthodontic force system it must be assumed that all aspects of the appliance are at rest. Therefore, the sum of the moments and forces acting on the appliance must be equal to zero. This assumption makes it possible to calculate the applied force system when there is adequate knowledge of some aspects of the system. Because static equilibrium states that the sum of all forces equals zero ($\Sigma F=0$) and the sum of all moments equals zero ($\Sigma M=0$), we can use formulations to calculate unknown force values when certain values are known. For instance, if a mesial force of 100 grams is applied to a canine bracket the tooth will tip until acted upon by another force. In this case it occurs when the bracket engages the archwire. If we know the distance from the bracket to the center of resistance of the canine is 10mm and the bracket width is 4mm we can calculate the normal forces of the couple acting within the bracket slot (See Figure 8).

<u>Example Calculation</u>:

Static Equilibrium: $\Sigma F=0$; $\Sigma M=0$

M_F (Moment of Force) = 100 grams x 10mm = 1000g-mm

M_C (Moment of Couple) = F_C (Normal Force of Couple) x 4mm

$\Sigma M=0$, therefore $M_F + M_C = 0$

$$1000\text{g-mm} + 4\text{mm}(F_C) = 0$$

$$4\text{mm}(F_C) = -1000\text{g-mm}$$

$$F_C = -250 \text{ grams}$$

Therefore, we know from this calculation that the archwire must be able to deliver 250 grams of force at the bracket archwire interface in order to offset the moment of the external force of 100 grams. If we understand the force system produced by the

appliance in equilibrium we can more accurately predict the treatment response. This understanding allows us to plan for the desired tooth movements, as well as make compensations for the undesirable side effects.

The Fixed Orthodontic Appliance:

The orthodontic appliance we know today was introduced by Edward Angle in the late 1920's. Angle realized that by turning his Ribbon Arch appliance "edgewise", orienting the long side of the rectangular archwire buccolingually, he could gain greater control over root position [9]. This Edgewise appliance was the first to use rectangular archwires and a rectangular bracket slot. Angle's Edgewise appliance improved the ability of the orthodontist to control tooth movement, but still required multiple archwire bends to achieve ideal alignment, root angulation and torque.

In the 1980's, Larry Andrews introduced the Straight-wire appliance. Andrews' appliance, by incorporating off-sets, root tip and torque in the bracket itself, reduced the need for the excessive wire bending required by the Edgewise appliance [10]. Though Andrews' original prescription has been modified by many, the Straight-wire appliance continues to be the instrument of choice for orthodontists. Due to many modifications of Andrews' bracket prescription, this appliance is generically referred to as a preadjusted appliance.

The typical preadjusted orthodontic appliance consists of three components; brackets, archwires and ligatures. In the case of self ligating brackets, a ligature is not necessary because a ligation mechanism is incorporated into the bracket face by the manufacturer. Because of the role that couples play in controlling tooth movement, it is the interaction of these components of the appliance that determine how the tooth will move relative to its three planes of space. As in the original Edgewise appliance, the bracket's rectangular slot allows for two points of contact in both the buccolingual and mesiodistal planes when used in conjunction with a rectangular archwire. As mentioned previously, these two points of contact create a couple, which aids in the three dimensional control of tooth movement.

The archwire functions as a spring, storing energy to be delivered through the bracket to direct tooth movement. In general, the archwire is the primary source of stored energy, though auxiliary springs and elastics may also be used. Its physical properties dictate the magnitude and duration of force that will be delivered to the dentition. The ligature's roles are to keep the archwire fully seated in the bracket slot, to maintain the contact points between the bracket slot and archwire and to provide the buccolingual couple required for first order control. Ideally, the ligature will deflect and activate the archwire "spring" while limiting the dissipation of energy as it is transferred from the archwire to the tooth. Variations in any of these three components may alter the nature of the force system being applied to the supportive structures of the teeth.

According to Burstone, the optimal orthodontic appliance has three main characteristics: 1) Accurate control of the center of rotation during tooth movement. 2) Optimal stress levels in the periodontal ligament. 3) Constant level of stress during tooth movement [5]. The first, control of the center of rotation, is determined by the ratio of the moment to the force (M/F) [6,8,11]. The orthodontic appliance must provide a means for producing a couple in order to control for the moment of the force that is being applied at the bracket. As described previously, the relative magnitude of these two variables, the moment and the force, affects the location of the center of rotation. Tipping, translation and root movement are three types of tooth movement that can be achieved through variations in the M/F ratio [7].

The second characteristic of an orthodontic appliance is the ability to apply an optimal stress level in the PDL. This is best achieved by using materials with a low load-deflection rate. The load-deflection rate is a measurement of how a spring's resistance or force delivery changes over a distance of activation. A low load-deflection rate is most desirable for orthodontic tooth movement because the force level will remain relatively constant as the degree of deflection becomes less. For example, as crowded teeth begin to align, archwire deflections become less severe and force levels decrease. If force levels decreased significantly with a small change in wire activation, tooth movement would slow or stop until the archwire was changed and an adequate force level was regained. Instead, with a low load-deflection rate, force levels remain relatively constant and tooth movement continues as the teeth align.

Also, low load-deflection archwires allow for greater accuracy in controlling force magnitudes. A small error in the deflection of an archwire with a high load-deflection rate will result in a large increase or decrease in force magnitude. This could result in either improper tooth movement or unwanted side effects. Conversely, a low load-deflection archwire provides a larger range of deflection that will produce relatively small variations in the applied force.

The third important characteristic of an orthodontic appliance is the maximal elastic load, which is the greatest force that can be applied without causing permanent deformation. Tooth movements and archwire activations should occur well below this maximal elastic load. The greater the maximal elastic load, the greater the range the archwire may be deflected without risking deformation or breakage. When an appliance has a low maximal elastic load it is at greater risk for plastic deformation or breakage, which means there is a greater risk of inefficient and/or unwanted tooth movement [5].

Orthodontic Treatment Sequence

In orthodontics, tooth movement is defined relative to its three planes of space. Movements in these planes are categorized as first order, second order and third order. First order tooth movement, commonly thought of as "in/out", refers to movements that can be viewed from the occlusal perspective. These include buccolingual or labiolingual movements, as well as rotations about the long axis (coronal-apical) of the tooth. Second order movements, referred to as "tipping", can be viewed from the buccolingual

perspective. These include movements in an occlusogingival direction or rotation about a buccolingual axis. Rotation about this axis would result in tipping of the root or crown in a mesial or distal direction. Finally, third order movement, commonly thought of as "torque", can be viewed from a mesiodistal perspective or buccolingual crossection. Third order refers to movement about a mesiodistal axis. This movement is important when attempting to achieve proper incisor inclination.

These three types of tooth movement are achieved through interaction of the archwire with the bracket slot and ligature. In the case of first order movement, the couple is generated when two points of contact occur between the bracket slot and archwire and/or the archwire and ligature. Because the force of ligation occurs only in a buccolingual direction, second order and third order movements are limited to interactions between the bracket slot and archwire (See Figure 12). Second order couples are formed when the archwire engages the bracket slot at a mesial and distal contact point, while third order couples have contact points that are oriented buccal and lingual.

Orthodontic treatment typically follows a sequence of correction through these three orders of tooth movement. The initial phase of orthodontic treatment, alignment, focuses on first order movements. During this phase, archwires with low load-deflection rates and large elastic ranges are used because they allow for the delivery of low constant forces over a wide range of movement. To deliver an adequate force system, the clinician selects an archwire dimension and material that can be easily deflected, fully

ligated in the bracket slot and delivers a low constant force over a large range of activation.

The second phase involves second order movements such as leveling and root tip. Leveling, or occlusogingival movements, can be accomplished using the force systems generated during the initial phase, but root tip requires increased stiffness of the archwire in order to deliver an adequate applied moment. To achieve the necessary intrabracket couple, the archwire must be of adequate dimension to engage the mesial and distal corners of the bracket slot. Once the necessary dimension is selected the clinician can dictate the force level and load-deflection rate by varying the archwire material.

The final phase of treatment, third order correction, involves the full expression of the prescription determined by the preadjusted appliance. To fully express the prescription and engage the third order couple, the archwire's dimension must approximate that of the bracket slot. By filling the bracket slot with a full size archwire, the clinician can assure that all couples will be engaged and any remaining correction will be resolved. Also, because the buccolingual dimension of the bracket, 0.028", is approximately four times smaller than the mesiodistal dimension, 0.138", the archwire must be of sufficient stiffness to deliver the higher force values necessary for an adequate third order couple (See Figure 12).

Orthodontic Archwires

Archwire Properties

There are many properties of archwires that are to be considered when selecting one for treatment. As previously discussed, the phase of treatment and order of tooth movement will dictate the desirability of certain properties. Therefore, the clinician should understand how variations in archwire material and dimension can affect the efficiency and performance of the preadjusted appliance.

Primarily, clinicians use three different wire materials during the course of orthodontic treatment. They are stainless steel, nickel titanium and beta titanium. These three materials are available in all dimensions of orthodontic archwires, including round, square and rectangular crossections. These wires can vary in dimension from 0.012" diameter round wire to 0.022" x 0.028" rectangular wire. Though they are available in the same dimensions and crossections, stainless steel, nickel titanium and beta titanium each has unique physical properties that make it better suited for specific phases of treatment.

Several factors are important when choosing an archwire. Kapila defined seven characteristics that he considered to have the most clinical relevance. These characteristics are: 1) Springback or working range 2) Stiffness or load deflection rate 3) Formability 4) Modulus of resilience or stored energy 5) Biocompatibility 6) Joinability and 7) Friction [12].

Springback

Springback, or working range, is related to the ratio of the yield strength to the modulus of elasticity (YS/E). A greater working range allows for larger activations, resulting in an increased working time for the appliance. It is also a measure of the elastic limit, or how far a wire can be deflected before experiencing permanent deformation [12](See Figure 13). Nickel titanium has physical properties that provide it the largest working range of the three materials [13]. When compared to stainless steel, beta titanium wire can be deflected twice as much without permanently deforming [14].

Load Deflection

Another characteristic is the load deflection rate, or stiffness of the wire. This is the force delivered by an appliance over a given range of deflection, and is proportional to its modulus of elasticity [12,15]. The modulus of elasticity is determined by calculating the slope of the elastic region of the load deflection curve (See Figure 13). A low load deflection rate is beneficial for the early phases of orthodontic treatment because it provides lower forces, more constant forces and greater accuracy in the application of force. Nickel titanium wires display the lowest load deflection rates, followed by beta titanium and then stainless steel [13,16,17].

Modulus of Resilience

The modulus of resilience is the amount of energy stored in the wire. It corresponds to the area under the curve for the elastic range of the wire. This is an important value

because it determines the amount of work available to move teeth. When the bending angle or torsional moment is the same for the three wire materials, the nickel titanium wire shows the greatest amount of stored energy, followed by beta titanium and then stainless steel (See Figure 14). This increased energy should result in increased clinical efficiency by decreasing the number of archwire changes and therefore the number of visits [13,16].

The Nickel Titanium Archwire

Nickel titanium alloys were first introduced to orthodontics by Andreasen in 1971 [18]. This alloy, referred to as Nitinol, was developed by Buehler at the Naval Ordinance Lab in the early 1960's. It was the first "shape memory" alloy and gained wide acceptance in orthodontics due to its high springback and low stiffness [16] (See Figure 15). In 1985, Burstone introduced Chinese NiTi, another nickel titanium alloy. This new alloy had 1.6 times the springback, 36% the stiffness and less permanent deformation when compared to Nitinol [17] (See Figure 16).

Physical Properties of Nickel Titanium

Ideally, a nickel titanium wire should retain its shape at intraoral temperature but be formable at a lower room temperature. This property gives the clinician the ability to ligate the archwire into the bracket with relatively little effort, but have it regain shape upon unloading at higher intraoral temperatures. During this shape regaining, the archwire is applying light, continuous force to the dentoalveolar structures. According to

Andreasen and Morrow, nickel titanium provides many advantages for the patient and the clinician. These advantages include fewer wire changes, less chairside time, reduced time to align and level and less patient discomfort [16]. The properties of nickel titanium alloys responsible for these clinical advantages are shape memory, superelasticity and hysteresis.

Shape Memory

Andreasen and Morrow described "Shape memory" as the ability of a wire to return to a previously manufactured shape when heated through a transition temperature range (TTR) [16] (See Figure 17A). Below its TTR the wire can be significantly deformed, but once heated the wire quickly returns to its original manufactured form. At low temperatures, below the martensitic finish temperature (M_f), nickel titanium alloys exist only in the martensitic phase. As the temperature increases, the alloy begins to convert from martensite into austenite. Once the temperature surpasses the austenitic finish temperature (A_f) for the alloy, the material has completely transformed into its austenitic phase and returned to its programmed shape. In the absence of localized mechanical stress, an archwire that is 100% austenitic is no longer superelastic.

The change from a deformed shape to the "memorized" form involves a transformation from the martensitic phase to the austenitic phase. This phase change occurs as an actual shift in the crystal structure of the alloy without changing the molecular structure. In order for this shape memory effect to occur intraorally, the alloy's austenitic finish (A_f) temperature needs to be set below oral temperatures. This is because the alloy is

manufactured to regain shape only once it has surpassed its A_f. The nickel titanium alloy loses much of its superelasticity when it exists purely in the austenitic form [19]. Therefore, the shape memory property of nickel titanium is not correlated with its superelasticity, but instead is another outcome of the alloy's uniquely flexible crystalline structure.

Superelasticity

Superelasticity, or pseudoelasticity, refers to nickel titanium's ability to return to its original form upon unloading after a significant deflection. This property is related to the transformation between distinct crystal structures. Nickel titanium can exist in three forms: martensite, austenite and stress-induced martensite. In response to temperature or mechanical stress the crystal structure is modified without a change in the atomic composition. Unlike that seen in other metal alloys, this stress related transformation from austenite to martensite is reversible. In the martensitic phase the alloy displays decreased stiffness, decreased springback and increased formability. The austenitic properties vary from those of the martensitic phase by providing increased stiffness, increased springback and reduced formability. Stress-induced martensite displays the superelastic properties, with a low modulus of elasticity and large working range (See Figure 17B).

Phase transformation of the nickel titanium alloy can occur in response to both temperature and mechanical stress. As an archwire is deflected the alloy undergoes a local phase transformation from austenite to martensite only at the point of high stress bending. This transformed phase is referred to as stress-induced martensite. Even if the

alloy is at a temperature above its A_f, localized stress can induce the phase transformation to occur. The local transformation to stress-induced martensite results in decreased stiffness and increased formability, therefore allowing the clinician to easily overcome large deflections during archwire ligation.

It is between the M_f and A_f that the superelastic properties of nickel titanium alloys are most apparent. This atomic equilibrium gives the crystal lattice a better ability to absorb stress. This results in a reduction of the modulus of elasticity, which means a lower load deflection rate. As a general rule, the austenitic phase of a superelastic wire will be stiffer than the martensitic phase, but both will be stiffer than a superelastic wire in phase transformation [19].

Hysteresis

The temperature required to transform martensite to austenite is slightly higher than that to transform from austenite back to martensite (See Figure 17A). This difference in transformation temperatures is referred to as hysteresis. Hysteresis also occurs during the stress-induced transformation from austenite to martensite [20] (See Figure 18). This means that the force required to deflect and ligate the archwire is greater than the force being delivered to the dentition upon unloading. Reducing the stress hysteresis means reducing the differential between the loading and unloading forces of the archwire. It has been suggested by Liaw et al. that reducing the stress hysteresis of an archwire could lead to a more favorable delivery of force in the clinical situation [21].

Orthodontic Archwire Ligation

According to Harradine, an ideal ligation system will have the following properties: 1) Be secure and robust. 2) Ensure full engagement of the archwire. 3) Be quick and easy to use. 4) Assist good oral hygiene. 5) Be comfortable for the patient. 6) Allow the clinician to incorporate high and low friction as desired. 7) Permit easy attachment of auxiliaries [22]. These properties focus on decreasing chairside and overall treatment time, improving patient acceptance and efficiently transmitting forces from archwire alloys.

Elastomeric and Wire Ligatures

Until the introduction of elastomeric modules in the 1970's, individual wire ligatures were routinely used to ligate archwires. These steel ligatures required that the clinician place, tighten, cut and then cinch the ligature on every tooth. Due to this time consuming process, elastomerics quickly replaced wire ligatures as the method of choice. Elastomerics are much faster and easier to place and more comfortable for the patient. They can also be fashioned into chains for space closure, as well as manufactured in various colors to improve patient acceptance. In relation to wire ligatures, the disadvantages of elastomerics include increased microbial colonization, incomplete seating of archwires for rotational control, rapid force decay, permanent deformation and increased resistance to sliding [23,24].

Taloumis reported that elastomerics lose 53% to 68% of their ligation force within the first 24 hours of activation. He also found that a simulated oral environment, one that is

warm and moist, lead to a more pronounced force decay and permanent deformation of the activated elastomeric [24] (See Figures 19-20). The uptake of water in the oral environment facilitates the slippage of molecules and increases the rate of force decay [25].

Self-Ligating Brackets

Due to these shortcomings of elastomerics, orthodontists have sought a ligation method that is as fast, easy and comfortable as elastomerics, but has the force constancy, improved cleansability and lack of deformation of wire ligatures. These brackets have become known as self-ligating brackets and have gained considerable popularity over the past fifteen years. It has been suggested that self-ligating brackets reduce chairside time, allow for longer appointment intervals due to consistent wire activation, provide better access for patient hygiene and reduce overall treatment time [26,27]. The reported disadvantages of self-ligating brackets include increased cost, reduced control of tooth movement, increased buccolingual profile, opening or breakage of ligation mechanisms and a tendency for archwire slippage resulting in tissue trauma [22].

Though they were first introduced to orthodontics in the 1930's, self-ligating brackets did not gain acceptance until the 1980's. Due to high manufacturing costs and poor acceptance, the early designs were quickly abandoned. Not until the early 1970's did self-ligating brackets return to orthodontics. The Edgelok® bracket was the first to be commercially available on a wide scale. The Edgelok® was a passive self-ligating bracket, but only achieved limited acceptance among orthodontists. In the early 1980's, another self-ligating bracket was introduced that incorporated an active ligation

mechanism. This bracket, called SPEED®, used a spring clip that interacted with the archwire. Since then many different self-ligating brackets have come onto the market. Commercially available self-ligating brackets can be differentiated as having either an active or passive ligation mechanism.

Active and Passive Ligation

Active SLBs (ASLBs) incorporate a clip that invades the bracket slot. This clip is able to flex and therefore can apply a seating force to the archwire. The force applied to the archwire varies with the archwire dimension and the first-order position of the tooth. Because the clip is not in constant contact with the archwire, these mechanisms have been referred to as interactive.[28] This is to differentiate them from conventional ligation brackets (CLBs) that use elastomeric and wire ligatures to provide a constant seating force, independent of wire size or bracket position. Examples of ASLBs currently on the market are the In-Ovation R® (GAC Intl), SPEED® (Strite) and Time® (American Orthodontics) brackets (See Figure 21).

Passive self-ligating brackets (PSLBs) have a rigid component that entraps the wire in the bracket slot without applying a force. With PSLBs, the archwire only engages the ligation mechanism when the wire is of adequate buccolingual dimension or the tooth is sufficiently malpositioned in the first-order.[28-33] Examples of PSLBs currently on the market are the Damon Mx® (Ormco), Carriere® (Class One) and the SmartClip® (3M Unitek) brackets (See Figure 22).

Friction in Orthodontics

Definition of Friction

Friction is defined as the force that resists movement of one surface past another and acts in a direction opposite the direction of movement[34]. There are two main components that influence the magnitude of the force of friction (F_{FR}). The first is the normal force (F_N), which is the perpendicular force responsible for holding the two surfaces in contact. The second is the coefficient of friction (μ), which is an experimentally derived constant dependent upon the surface characteristics of the materials in contact. The coefficient of friction exists as both a static (μ_s) and kinetic (μ_k) value. The static coefficient (μ_s) pertains when calculating the force required to initiate movement, while the kinetic coefficient (μ_k) is used once the surfaces are in motion. The coefficient of friction is calculated by dividing the force of friction (F_{FR}) by the normal force (F_N). Hence, the force of friction is described mathematically by the equation $F_{FR} = F_N \mu$.

In orthodontics, the moving surfaces in contact are the brackets, archwires and ligatures. These components of the fixed appliance initiate tooth movement by transmitting force from the archwire to the periodontal support of the tooth. As the tooth moves, the bracket, archwire and ligature are intermittently contacting and sliding against one another. When using a conventional preadjusted appliance, these surfaces must move past each other in order for tooth movement to occur. Because friction acts to dissipate the magnitude of force transmitted from the appliance to the teeth, many studies have been designed to measure and better understand the role that friction plays in

orthodontics and how variations in the three appliance components affect the delivery of force. Most studies have used in vitro models to vary the components of the fixed appliance. The brackets, archwires and ligatures have been varied in the following ways: material, dimension, ligation method, bracket angulation, relative motion and biological resistance. From these studies we have learned much about the interaction of these components and how each plays a role in determining the magnitude of frictional resistance.

Material Composition

The influence of material composition on friction can be divided into two categories: surface characteristics and mechanical properties. Angolkar, when sliding an archwire through parallel bracket slots, reported that ceramic brackets exhibit more resistance to sliding than stainless steel brackets [35]. Angolkar and Bednar both propose this is due to the increased surface irregularities of ceramics. Magnification shows the ceramic surface to have a highly variegated topography, which leads to a higher coefficient of friction and greater frictional resistance when compared to stainless steel [29,35-38].

With regards to archwire material, most authors agree that stainless steel archwires generate less frictional resistance when compared to nickel titanium and beta titanium, with the beta titanium alloy showing the highest levels of resistance [31,36,39-42]. Beta titanium, at 80% titanium, has a higher coefficient of friction due to a more reactive surface chemistry than nickel titanium at 50% titanium. Kusy reported that the higher the percent composition of titanium the greater the surface reactivity, and subsequently the

greater the frictional resistance [43]. Tidy's study of sliding frictional resistance found that nickel titanium wires have twice and beta titanium five times the frictional resistance of stainless steel [42].

Conversely, some investigators have shown variations in frictional resistance when alloys are subjected to second order binding forces [44,45]. Frank and Nikolai's findings showed nickel titanium to have less frictional resistance than stainless steel when second order angulation exceeded a certain limit [44]. This limit of angulation was defined by Kusy and Whitley in 1999 as the critical contact angle. It refers to the degree of bracket angulation required for the wire to engage mesial and distal contact points within the bracket slot [46] (See Figure 26). When second order angulations exceeded this critical contact angle, binding and increased frictional values were seen. Both Frank and Nikolai [44] and Thorstenson and Kusy [45] reported that the nickel titanium alloy showed less frictional resistance than the stiffer stainless steel. Frank and Nikolai proposed that this was due to nickel titanium's low load deflection rate, which results in a lower normal force within the bracket slot, and thus a lower force of friction [44].
Though the frictional resistance was found to be less with nickel titanium archwires, this should not be taken to mean that nickel titanium is the preferred material during sliding mechanics. In the clinical situation, a reduced resistance to sliding is believed to be preferential, but an adequately stiff archwire that provides the uprighting moment required for controlled tooth movement is significantly more important.

Material Dimension

Multiple studies have assessed the effect of varying the size and shape of brackets and archwires on frictional resistance. When considering archwire shape and dimension, most authors agree that friction increases with the cross-sectional area of the archwire [29,35,40,41,47,48]. Some researchers have reported a deviation from these findings, which they propose is a result of wire notching during increased second order angulation [44,46].

Frank and Nikolai described this notching phenomenon by considering variations in pressure at the contact points between the archwire and bracket. Pressure is calculated by dividing the applied force by the surface area of application, $P = F/A$. Frank and Nikolai proposed that round wires will contact the bracket slot at a single point resulting in a higher pressure due to the decreased surface area. Conversely, rectangular wires contact the bracket slot over a line, which increases the surface area in contact and therefore decreases the pressure. It is suggested that high pressure at a point contact can result in increased notching of the round wires and a subsequent increase in resistance to sliding [44].

Bracket width has been shown to affect frictional resistance in multiple ways, but there is controversy as to whether it increases or decreases friction. Some authors suggest that increased bracket width decreases the force of friction by decreasing the interbracket distance. This results in increased stiffness of the wire segment, which leads to decreased bending and a decreased risk of binding[29,40,42]. Conversely, a stiffer wire segment would mean a greater load for an equivalent degree of deflection. This greater load would result

in larger normal forces within the bracket slot, which could translate into an increase in frictional resistance.

Also, the geometry of the bracket slot shows that second order binding occurs at smaller angulations for wider brackets when comparing them to more narrow ones. This means that wider brackets require less tipping, in either first or second order dimensions, for the archwire to engage the slot and generate a couple. Frank and Nikolai suggested that this earlier binding resulted in an increased normal force and therefore an increased frictional resistance for wider brackets[44]. Kapila et al. and Ogata et al. both suggest that increased bracket width increases friction by increasing the activation force necessary to place an elastomeric ligature. This increased elastomeric activation results in an increased normal force and therefore increased force of friction [24,41,49].

Bracket Ligation

Due to its direct effect on the normal force, the method of ligation may have the most significant effect on frictional resistance. Ligation techniques can be divided into three categories: active, interactive and passive. Active ligatures have mechanical properties that allow them to apply a force to the archwire in order to hold it in the bracket slot. Active ligatures may be elastomeric rings or stainless steel ligatures. Interactive ligation, as defined by Voudouris, consists of a precision arm that extends into the bracket slot[28]. The precision arm interacts and applies a force to the archwire under two circumstances: (1) the archwire is of significant buccolingual dimension to contact the arm or (2) the bracketed tooth is sufficiently malpositioned. Passive ligatures function by converting

the bracket slot to a tube. The passive ligature restrains the archwire within the bracket slot, but the ligature has no mechanical properties that allow it to function as a spring or apply a force to the archwire.

Due to their ease of placement, low cost and patient acceptance, elastomerics have been the ligature of choice for many years. As previously mentioned, the disadvantages of elastomerics are stress relaxation, bacterial plaque accumulation and breakdown in the oral environment [24,25,50]. Stainless steel ligatures provide a more constant force and reduced plaque accumulation, but they require more time to place and their ligation force can vary significantly depending on the clinician's technique [23,24,51]. Whether using elastomeric or stainless steel ligatures, it has been estimated that the force of ligation can range from 50 to 300 grams[40,44,47,52].

In the case of self-ligating brackets, the ligation force varies with the type of mechanism and the position of the bracketed tooth. When teeth are ideally aligned, passive systems apply no ligation force, while the ligation force of active and interactive systems varies with the archwire dimension. In the presence of malaligned teeth, ligation force will vary with the degree of malposition, the ligation mechanism and the mechanical properties of the ligated archwire.

The equation to calculate the force of friction, $\mathbf{F_{FR} = F_N\, \mu}$, shows that an increase in the normal force (F_N), in our case the force of ligation, results in a proportional increase in frictional resistance (F_{FR}). Therefore, any ligation technique that applies a force to the

archwire should inherently produce more frictional resistance. When comparing the frictional properties of active, interactive and passive ligation methods, studies generally support this relationship.

Henao and Kusy, using a pretreatment typodont model, reported that passive self-ligating brackets produced less frictional resistance when compared to conventional active ligation. While a significant difference was seen during sliding of small diameter archwires, Henao and Kusy showed no statistical significance as the wire size increased to engage the self-ligating mechanism[53,54]. Voudouris and Pizzoni et al., using a parallel fixed bracket model, both showed PSLBs produced significantly less frictional resistance than both CLBs and ASLBs. They also confirmed that self-ligating brackets in general reduced sliding frictional resistance when compared to conventional ligation techniques [28,31].

The methods of these studies must be questioned when evaluating the authors' conclusions. The Henao and Kusy studies measured friction by pulling a ligated archwire through brackets bonded to a pre-treatment typodont. Though this test measures friction related to a wire moving through brackets, what does it suggest of the clinical situation and the ability of the various ligation mechanisms to efficiently move teeth? There is an assumption that a reduction in friction translates into more efficient tooth movement, but as discussed previously, controlled tooth movement requires a force and an applied moment. Obviously, there is more play present in a passive ligation mechanism, which translates into a lower normal force and reduced resistance to sliding. The failure of the

ligature to fully seat and activate the archwire may reduce friction, but it also reduces the ability of the archwire to provide an adequate force system for efficient tooth movement. Pulling an archwire through ligated brackets allows us to measure frictional resistance, but any assumptions regarding the clinical implications of these findings would be misleading.

Clinical Studies

In 2005 and 2006 Miles reported two prospective clinical trials that used a split mouth design to compare the effectiveness of passive self-ligating brackets to conventional twin brackets. A baseline irregularity index was determined for each patient's mandibular arch and then remeasured at each of two 10 week intervals. A 0.014" Damon copper NiTi wire (Ormco) was activated for the first ten weeks and then replaced by a 0.016" x 0.025" Damon copper NiTi (Ormco) for the second ten weeks of alignment.

Miles reported that neither the Damon 2 nor the SmartClip passive self-ligating brackets were more effective at reducing irregularity than the conventional twin bracket with elastomeric ligation. In the Damon 2 study the conventional twin bracket achieved a lower irregularity index at both 10 and 20 weeks of alignment, but this value was not statistically significant [55]. In the SmartClip study the conventional brackets showed a statistically significant reduction in irregularity at 10 weeks, but at 20 weeks there was no significant difference between the two groups [56].

For the Damon 2 study Miles also measured patient comfort with the appliances. During initial alignment the Damon 2 bracket was less painful, but when the larger archwire was ligated at 10 weeks it was found to be more painful that the conventional ligation. This is likely due to remaining irregularities requiring significant deflection of a rectangular archwire. Due to its greater cross-sectional area, the rectangular archwire exhibits a greater unloading force. Because the passive ligation mechanism does not allow for any flexure or force dissipation, the patient feels more force and therefore more discomfort following ligation [55].

In another prospective study reported in 2007, Miles compared the rates of tooth movement with SmartClip passive self-ligating brackets to conventional twin brackets during en masse space closure. All anterior teeth were bonded with the same conventional twin brackets. Each patient had SmartClip brackets bonded to second premolars and molars on one side and conventional twin brackets on the other. The patient was blinded according to the type of ligation. Space closure was accomplished with a 0.016" x 0.022" stainless steel archwire, nickel titanium coil springs and stainless steel ligation of the posterior conventional twin brackets.

Miles showed no significant difference in the rate of space closure between the passive self-ligating and conventional twin tied with stainless steel ligatures. Teeth on the SmartClip side moved at a median rate of 1.1mm per month, while teeth on the conventional side moved 1.2mm per month. This study suggests that passive ligation

may not provide a significant clinical advantage over conventional ligation during en masse space closure [57].

These clinical studies imply that friction may not be an overwhelming factor with regard to the efficiency of tooth movement. In fact, Braun suggests that the "jiggling" effect of teeth within the PDL renders friction inconsequential in the clinical setting.[58] Therefore, there may be other factors influencing the bracket-archwire interaction that may affect treatment efficiency. We hypothesize that the interaction of the archwire and ligation mechanism affects force delivery in the first order, and that understanding how archwire selection can influence force delivery can enhance treatment efficiency.

Hypothesis

1. Degree of bracket play varies with the ligation mechanism and archwire diameter.
    a. PSLBs exhibit greater bracket play than ASLBs and CLBs.

2. Deactivation force varies with the ligation mechanism and archwire diameter.
    a. Using a Cantilever Test, CLBs and ASLBs will deliver higher deactivation forces than PSLBs because of the active nature of the ligation mechanism.
    b. Using a Three Point Bend Test, SLBs will return higher deactivation forces than CLBs because SLBs will reduce frictional resistance and result in less dissipation of the deactivation force.

3. Superelastic range varies with the ligation mechanism and archwire diameter.

Objectives

1. To measure the degree of archwire deflection required for the ligation mechanism to engage the first order couple when bracket type and wire diameter are varied.

2. To measure the force of deactivation produced by CLBs, ASLBs and PSLBs while varying the degree of first order deflection and archwire diameter.

3. To examine the effect of ligation on the superelastic range by varying bracket type, archwire diameter and the degree of deflection.

Materials and Methods

Brackets and Wires Tested

We evaluated five different orthodontic brackets: Victory™ (3M Unitek, St. Paul, MN), SmartClip™ (3M Unitek, St. Paul, MN), InOvation-R™ (GAC Intl., Bohemia, NY), Damon MX™ (Ormco, Glendora, CA), and Carriere™ (Class One, Lubbock, TX). We chose these brackets because each uses a different method for ligating the archwire into the bracket slot. For ease of study we chose to limit our testing to maxillary premolar brackets with an occlusogingival slot dimension of 0.022". See Table 1 for a description of the various brackets' materials, dimensions and ligation mechanisms. In combination with these orthodontic brackets, four superelastic nickel titanium archwire segments were tested. The samples tested varied only in diameter and were obtained from the same manufacturer (Ultimate Wireforms, Inc., Bristol, CT). See Table 2 for a list of the archwires tested.

Table I: Description of Brackets Tested

Bracket Type	Ligation Mechanism	Slot Width	Effective Width*	Material	Manufacturer
Victory™	Conventional Elastomeric	2.98mm	3.73mm	Stainless Steel	3M Unitek
SmartClip™	Passive Clips	4.15mm	3.80mm	Stainless Steel NiTi Clips	3M Unitek
Damon MX™	Passive Slide	2.67mm	2.39mm	Stainless Steel	Ormco
Carriere™	Passive Slide	3.35mm	2.84mm	Stainless Steel	Ortho Organizers
In-Ovation R™	Interactive Clip	2.97mm	2.97mm	Stainless Steel Elgiloy Door	GAC Intl.

* Distance between the contact points formed during a first-order couple. Varies with the width of the slot base and the ligation mechanism.

Table II: Dimensions of Nickel Titanium Archwires Tested

Superelastic NiTi TTR = 55° F	0.014"	0.016"	0.018"	0.020"	Ultimate Wireforms Inc., Bristol, CT

Testing Apparatus

This study consisted of two different testing methods. Each of these methods used the same custom bracket fixture fabricated specifically for this study. The custom fixture consists of an aluminum base and two stainless steel stanchions placed 10mm apart. Each of these stanchions has a vise-like mechanism capable of holding a bracket in place by applying pressure to the bracket base (See Figure 23). The fixture was designed to be mounted to the crosshead of a Tinius Olsen H1-KS mechanical testing machine. This mechanical testing machine incorporates a vertical load cell with a 50N maximum load and is encased in a Plexiglas cabinet for temperature control (See Figure 24). The temperature inside the cabinet was maintained using a temperature sensor and heated blower. All tests were performed at 37 degrees Celsius to mimic intraoral temperature.

Calibration

For all experiments the fixture was initially calibrated to the center of the crosshead using a 0.071" gauge. The brackets were then mounted into the fixture, paralleled with a 0.021" x 0.025" stainless steel wire segment and the interbracket distance was set to 10mm using a caliper. The interbracket distance was measured between the termination points of the bracket slots, not from the bracket midpoint (See Figure 23). Once the fixture was calibrated and the brackets were mounted, the fixture was attached to the

crosshead of the testing machine and approximated with the load cell carriage. A 0.021" x 0.025" stainless steel wire segment was used to determine the zero position of the crosshead. Zero position was defined as passive contact between the arch wire segment and the load cell carriage. The tolerance for passivity was when the carriage and archwire were in contact but measured less than 2 grams of force. Once the zero position for the crosshead was set, testing of the brackets and archwire segments could begin.

Cantilevered Archwire Test

The first test uses a cantilever design to measure the interaction between the archwire and bracket ligation mechanism. This design was chosen to limit the influence of sliding frictional resistance on the measurement of the loading and unloading forces. A 10mm straight segment of archwire obtained from the visually straight posterior segments of a standard arch form is ligated in one bracket using the prescribed method. In the case of the Victory bracket we used silver elastomeric rings from Class One Orthodontics (#185-58) and Mathieu pliers to place the elastomeric ligature. The archwire segment extended 5mm from the termination of the bracket slot to rest on the carriage of the load cell. Each test consisted of the loading and unloading of the archwire segment in the buccal direction over a range of 4mm at a rate of 10mm/min.

Three Point Bending Test

The second test uses a three point bend design, similar to that commonly used for testing the mechanical properties of orthodontic wires. In this test, the archwire segment is ligated into both brackets, which are fixed at an interbracket distance of 10mm (See

Figure 23). The fixture and brackets are calibrated and mounted as previously described in the cantilever test. Once the zero position is defined, a 20mm straight archwire segment obtained from the visually straight posterior segments of a standard arch form is ligated into both brackets using the prescribed ligation method. As in the cantilever study, we used silver elastomeric rings from Class One Orthodontics (#185-58) and Mathieu pliers to ligate the archwire into the Victory bracket. The loading and unloading of the wire occurred at the midpoint of the ligated brackets over a range of 4mm at a rate of 10mm/min.

Data Acquisition and Analysis

All combinations of brackets and archwires were tested three times with a new wire segment used for each test. In the case of the Victory group, a new elastomeric ligature was placed for each wire sample tested. Data was collected continuously at a sampling rate of 20Hz in the form of ASCII files. These files were transferred to a Microsoft Excel spreadsheet for processing and analysis.

Bracket play was defined as the amount of wire deflection required to initiate the first-order couple. The degree of deflection required for bracket/archwire contact was measured at a threshold of 5 grams of loading force. The activation and deactivation forces were measured continuously and the means ± SD were calculated at 1mm, 2mm and 3mm of deactivation. The values for superelastic range were determined graphically by selecting the end point for the superelastic plateau upon deactivation. The magnitude of deflection at the end point was subtracted from the total 4mm deflection to calculate

the superelastic range. Data was analyzed using a single factor ANOVA (α=0.01) with the bracket type as the discriminating variable. Group differences were further analyzed using post-hoc T-tests with α=0.01.

Results

Cantilever Test

Figure 25 demonstrates a typical load deflection curve. As a reference, bracket play, deactivation force and the superelastic range have been identified. Complete load-deflection curves for the five brackets tested are seen in Figure 26A-D. The load-deflection curves show that ligation type and archwire diameter have a significant effect on the degree of bracket play, deactivation load and superelastic range.

Bracket Play

As seen in Table III, the Victory bracket engaged the first-order couple at 0.57 ± 0.004 degrees of wire deflection when paired with the 0.014" wire, followed by the In-Ovation R (8.13 ± 0.013 deg) and SmartClip (12.08 ± 0.010 deg) brackets. The Damon MX (14.66 ± 0.029 deg) and Carriere (14.79 ± 0.010 deg) brackets displayed the greatest degree of bracket play and were not significantly different from each other. As Figure 27 shows, trends remained intact during deflection of the 0.016", 0.018" and 0.020" archwires, but the Victory bracket showed increased play with the 0.020" wire deflection. The Damon MX and Carriere brackets were not significantly different for any wire diameter tested and consistently showed the greatest amount of bracket play

Deactivation Force

The deactivation force varied with the ligation mechanism, degree of deflection and archwire diameter. In Table IV, the mean loads at 1mm, 2mm and 3mm of deactivation

are shown for the various bracket-archwire combinations. The Victory bracket varied significantly from the SLBs for all archwire diameters at the 1mm deflection ($p \leq 0.01$). As the degree of deflection increased to 2mm and 3mm, force values continued to vary significantly between the Victory CLB and the other SLBs, but the mean force values approximated each other in magnitude (See Figures 29 and 30). For all wire diameters tested, PSLBs generally displayed lower deactivation forces when compared to the Victory and In-Ovation R brackets. The one exception was when the Victory bracket was paired with the 0.020" archwire. The 0.018" and 0.020" 3mm deflection showed the Damon MX (90.21 ± 1.66g, 148.07 ± 2.43g) and Carriere (88.83 ± 1.03 g, 148.05 ± 0.77g) brackets to produce the lowest deactivation forces of all brackets tested (See Figure 30).

As seen in Figure 28, Victory and In-Ovation R were the only brackets to produce a measurable force with both archwires at 1mm of deactivation. The SmartClip bracket provided an unloading force with the 0.016" wire at 1mm of deflection, while the Damon MX and Carriere brackets produced no substantial force at 1mm until the diameter was increased to 0.020". For all archwire diameters and deflections tested, the Damon MX and Carriere brackets were not statistically different ($p \leq 0.01$).

<u>Superelastic Range</u>

As shown in Figure 31, the superelastic range was significantly different for all brackets except Damon MX and Carriere ($p \leq 0.01$). When coupled with the 0.014" archwire, the Damon MX (1.77 ± 0.02 mm) and Carriere (1.75 ±0.02 mm) brackets provided the

smallest superelastic range and were not significantly different for any archwire diameter tested. Table V lists the mean ± SD values for all brackets and archwires tested. The Victory (3.22 ± 0.04 mm, 3.18 ± 0.03mm, and 2.84 ± 0.01 mm) bracket had the largest superelastic range for the 0.014", 0.016" and 0.018" groups respectively. The superelastic ranges for all PSLBs approximated each other as the wire diameter increased. The In-Ovation R bracket had the largest superelastic range of all SLBs with the 0.014" wire (2.21 ± 0.01 mm), but the smallest when the wire diameter increased to 0.020" (2.06 ± 0.02 mm). The superelastic range for the Victory (0.99 ± 0.24mm) bracket fell significantly when wire diameter increased to 0.020", while the PSLBs showed an increasing superelastic range with increasing wire diameter.

Three Point Bend Test

Complete load-deflection curves for the five brackets tested are seen in Figure 32A-D. The SmartClip bracket was not able to withstand the activation force when the 0.018" and 0.020" wires were tested; therefore no data is available for these bracket-archwire combinations. The load-deflection curves show that ligation type and archwire diameter have a significant effect on the deactivation force. Because the three point bend test requires that the archwire slides through the brackets, frictional resistance will influence the measured forces of deactivation. All means ± SD with statistical comparisons can be found in Table VI.

1mm Deflection

Figure 33 shows a comparison of the four wire diameters across all bracket types. For the small diameter 0.014" wire, all SLBs exhibited similar forces levels and were significantly lower than the Victory (151.82 ± 2.41g) bracket. The SmartClip and In-Ovation R brackets were not statistically significant, nor were the Damon MX or Carriere different from each other. As wire diameter increased, the SLBs began to display greater variation. The SmartClip and In-Ovation R brackets showed no significant difference, but both were significantly different from the Damon MX and Carriere. The Victory bracket continued to deliver the greatest deactivation force.

As stated previously, the SmartClip bracket was unable to hold larger diameter wires during deflections of 3mm or greater. Therefore, no load-deflection data is available for this bracket with the 0.018" or 0.020" archwires. When wire diameter increased to 0.018", the In-Ovation R (188.55 ± 5.55g) bracket provided a deactivation force that was not significantly different from the Victory (181.76 ± 4.24g). The Damon MX and Carriere brackets continued to deliver lower deactivation forces with Damon MX (134.96 ± 2.30g) significantly less than Carriere (152.00 ± 6.47).

2mm Deflection

Figure 34 shows a comparison of all bracket types across the four wire diameters tested. When testing smaller diameter wires, 0.014" and 0.016", all brackets delivered similar force levels. Carriere (166.34 ± 3.86g) was the only bracket that was significantly different with the 0.014" test. As wire diameter increased to 0.016", the Victory (196.04

± 7.09g) and In-Ovation R (209.03 ± 7.61g) brackets delivered significantly lower deactivation forces than the PSLBs. When the large diameter wires were tested there was no statistically significant difference between the SLBs ($p \leq 0.01$). The Victory (209.63 ± 5.71g, 264.90 ± 9.63g) bracket returned significantly lower deactivation force than all SLBs when coupled with 0.018" and 0.020" wire diameters.

3mm Deflection

Figure 35 shows a comparison of all bracket types across the four wire diameters tested. The SmartClip (139.07 ± 3.52g) and Carriere (152.27 ± 6.12g) brackets delivered the greatest deactivation forces with the 0.014" deflection and were significantly different than all other brackets. There was no statistically significant difference found with deflection of the 0.016" wire, though the SmartClip (193.12 ± 7.34g) delivered the highest force and Damon MX (160.87 ± 15.27) delivered the lowest. When wire diameter increased to 0.018" and 0.020", the Damon MX and Carriere brackets demonstrated the lowest deactivation forces and were not significantly different from each other.

Discussion

First order load-deflection curves were used to examine conventional (CLB), active self-ligating (ASLB) and passive self-ligating (PSLB) bracket types. Cantilever and three point bend models were developed to assess the bracket-archwire interaction. The cantilever model assesses the pure force system occurring at the bracket by eliminating the influence of sliding frictional resistance when measuring the activation/deactivation load. The influence of friction is limited because the wire is not required to slide through the bracket slot during deflection. A buccal deflection was chosen because it produces a class II lever, which results in a greater force being applied to the ligation mechanism.

Conversely, the three point bend model requires that the wire slide through the brackets during loading and unloading. This two bracket model comes closer to simulating the clinical situation than the cantilever model, which was designed to focus specifically on the force system at the bracket. Because the wire must slide through the brackets, this test incorporates sliding frictional resistance into the load-deflection data. Results from both the single bracket cantilever model and the two bracket three point bend model show that the bracket type and archwire diameter have a significant effect on the degree of bracket play, force of deactivation and superelastic range.

Authors generally agree that SLBs reduce sliding frictional resistance when in passive alignment, but perform similarly to conventional ligation when archwire diameter and/or angulation are increased to engage the bracket.[44,45,49,53,54,59-61] Due to the concerns of

anchorage loss with increased friction during sliding mechanics[62,63], many studies have focused on varying second order angulation while testing stiffer archwire materials of greater cross-section. Until recently, little attention has been paid to the ligature-archwire interaction during first order movements[64,65]. Because the ligation mechanism is inherent in forming the first order couple that produces the axial moment, its interaction with the archwire in the first order will have a significant effect on the tooth moving force. Therefore, first order wire deflections should provide the best method for comparing the ligation mechanisms of various bracket types.

Our results show that CLBs and ASLBs significantly reduce the degree of deflection necessary to engage the first order couple, while narrow PSLBs like Damon MX and Carriere exhibit the greatest amount of bracket play. Due to their ligation slides being narrower than their slot bases, both Damon MX and Carriere brackets provide an effective first order width that is less than the actual bracket width (See Table I). The reduction in width results in a degree of first-order play that is greater than would be expected with geometric calculations. Typically, wider brackets will engage the first order couple earlier than more narrow brackets[66], but our data shows that the narrower In-Ovation R bracket demonstrates less bracket play than the wider SmartClip. This is likely a result of the interactive nature of the In-Ovation R ligation mechanism. Even though In-Ovation R is smaller mesiodistally than SmartClip and Carriere, its ligation clip engages the deflected archwire sooner.

When considering the correction of rotations, the degree of bracket play suggests how much correction may be accomplished with a given bracket-archwire combination. When tested with the larger 0.018" wire, the PSLBs provided a minimum of 9.13 degrees of play. This means that correcting a tooth that is rotated less than 9 degrees would require one of the following: a wire with increased buccolingual dimension, a wider bracket or addition of a more active ligation mechanism. The In-Ovation R bracket had only 8.13 ± 0.013 degrees of bracket play with the 0.014" wire, which is less than the PSLBs even when coupled with the larger 0.018" wire. The conventional Victory bracket showed the smallest degree of bracket play, but this test did not take into account the force decay of the elastomeric. Taloumis reported that elastomerics lose 53% to 68% of their ligation force in the first 24 hours of activation.[24] Therefore, bracket play can be expected to increase over time for the CLB. Although, the use of small diameter, low force wires with the CLBs would limit the risk of the deflected wire overwhelming the elastomeric.

Once the first-order couple is engaged, the PSLBs demonstrated lower deactivation forces compared to the Victory and In-Ovation R brackets. This is due to the passive nature of the ligation mechanism. As the deflected archwire is returning to its passive state, the unloading force of the archwire and force of the active ligation mechanism (elastomeric or self-ligating clip) combine to produce a deactivation force that is greater than that of the PSLBs.

Interestingly, no PSLBs were able to provide a measurable force with a 1mm deflection of the 0.014" wire, but both the Victory and In-Ovation R supplied a force level adequate for tooth movement. The SmartClip bracket was able to produce 36.96 ± 1.84 grams of force with the 0.018" wire, while the Damon MX and Carriere brackets required an 0.020" wire to produce a measurable force at 1mm of deflection. These findings suggest that PSLBs may require larger diameter archwires earlier in treatment in order to correct cases of minor crowding.

When examining the superelastic ranges for the various bracket types, we found that CLBs and ASLBs deliver greater superelastic ranges when compared to PSLBs. These findings agree with Hemingway et al. who found that PSLBs may not adequately strain the NiTi wire to take advantage of the superelastic effect.[67] The difference between bracket types becomes less apparent as wire diameter increases and bracket play decreases. For the 0.014" test, the Victory bracket provided 3.22 ± 0.040mm of superelastic activity over the full 4mm of deactivation. This was significantly greater than all SLBs tested. The In-Ovation R bracket had the greatest range for SLBs with the 0.014" wire, but this advantage was lost as wire diameter increased to 0.018". This is likely due to the flexible clip reducing stress on the deflected wire, which results in an earlier transformation back to the austenitic phase. When the wire increased to 0.018", the SmartClip (2.40 ± 0.025mm) bracket had the largest range of all SLBs. This is due to a larger bracket width combined with a passive ligation mechanism. The wider bracket provides reduced bracket play and therefore maintains the couple longer, while passive ligation maintains wire activation by remaining rigid in response to wire deflection.

The three point bend test data provided a different perspective regarding force delivery and archwire deflection. At 1mm of deflection all SLBs delivered a measurable force that increased and approached that of the CLBs as wire diameter increased. In fact, with large diameter wires, the SLBs provided significantly higher deactivation forces than the CLBs. This was likely due to a decrease in deactivation force for the CLBs. The increased normal force of ligation and coefficient of friction for the conventional elastomeric ligation resulted in increased sliding frictional resistance and a subsequent decrease in the measured deactivation force. These findings suggest that in cases of minor crowding the CLBs are best paired with small diameter wires and SLBs with larger diameter wires.

As the degree of three point deflection increased to 2mm, all brackets tested delivered similar deactivation forces for both the 0.014" and 0.016" wire diameters. This is likely due to two primary factors: 1) At 2mm of deflection all wires were adequately stressed and therefore transformed into their SIM superelastic phase. 2) The ligation force necessary to activate the wire is low enough so as not to overwhelm the elastomeric and result in increased sliding frictional resistance. As wire diameter increases, we see a similar reduction in deactivation force for the CLBs and increase in deactivation force for the SLBs that we saw with the 1mm deflection.

The 3mm three point deflection shows us that SLBs do experience sliding frictional resistance and that this effect is seen when larger diameter superelastic wires are used to correct severe first order deflections. When the smaller diameter wires, 0.014" and

0.016", are deflected 3mm all of the brackets deliver similar deactivation forces. When large diameter wires are tested, the PSLBs show a significant drop in the deactivation force when compared to the CLBs and ASLBs. This is likely due to binding frictional resistance occurring between the deflected wire and the rigid ligation mechanism. This "stick-slip" phenomenon can be clearly seen in Figures 32C and 32D. Notice how the far right of the deactivation curve becomes quite erratic for the SLBs. Also, as the degree of deflection lessens, the deactivation forces increase. This suggests that as deflection decreases so does the binding frictional resistance, which subsequently results in less dissipation of the deactivation force and greater return from the superelastic wire.

This observation can be explained by the dimension and design of the PSLBs. Because the Damon MX and Carriere brackets exhibit a narrower effective width and have a rigid ligation slide, the normal forces of ligation are significantly higher than the wider brackets with active mechanisms. These large normal forces between the bracket and wire result in large forces of friction (Friction = Normal Force x Coefficient of Friction). The CLBs and ASLBs have the ability to flex and therefore dissipate the ligation force, which results in reduced binding frictional resistance and greater deactivation forces.

It is apparent from these findings that the clinician must consider his or her bracket ligation method when selecting an archwire for the alignment phase of treatment. If using conventional elastomeric ligation, the clinician should consider small diameter, low force wires, such as 0.012" or 0.014" superelastic nickel titanium for the best results. Because conventional ligation allows for less than 1 degree of bracket play, large

diameter wires are unnecessary for adequate first order control. Also, larger wires require the elastomeric to provide greater ligation force to maintain the wire's activation, which could result in a loss of control and reduction in the superelastic range of the wire. Therefore, even with 50% force decay of the elastomeric, a small diameter NiTi wire would be more likely to maintain its superelasticity over the full range of activation.

If the clinician prefers the use of SLBs over conventional ligation, archwire selection should be based on the type of ligation mechanism being used and the width of the first order couple. As seen in this study, the active In-Ovation R bracket provided adequate first order control with the 0.014" wire at 1mm of deflection, while only the widest PSLB, SmartClip, was able to gain control when wire diameter was increased to 0.018". In this study, a 1mm deflection was equivalent to 11.3 degrees of axial rotation. This means that even with a 0.018" diameter wire, the Damon MX and Carriere brackets could not provide an adequate couple for correction. This suggests that clinicians using PSLBs need to consider bracket width and buccolingual archwire dimension when selecting the appropriate alignment wire.

It appears from our results that ASLBs provide more benefits than PSLBs during the alignment phase of treatment. The In-Ovation R bracket is narrower than both the SmartClip and Carriere brackets, but provides a superior reduction in bracket play and increase in superelastic range. Also, a narrower bracket will provide increased interbracket distance, resulting in a reduction in activation force of the wire. This means that the clinician can select a smaller diameter, lower force wire and expect it to work

over a greater range of activation. In order to gain similar control with PSLBs, the clinician must use a larger diameter wire, which would lead to an increased deactivation force and risk of increased sliding frictional resistance.

For instance, a 0.014" superelastic wire coupled with the In-Ovation R bracket will provide a maximum of 50.24 grams of deactivation force, while providing correction to within 8 degrees of rotation. A SmartClip bracket would require a 0.018" superelastic wire to provide correction to within 9 degrees of rotation, but this wire would produce a maximum of 93.39 grams of force. These increased force values with PSLBs might be remedied by using large diameter thermally activated nickel titanium wires. These wires could gain first order control earlier in treatment while delivering deactivation forces equivalent to smaller diameter austenitic wires.

The cantilever model allowed us to isolate forces occurring at the bracket, but did not take into account interbracket distance, deflection between multiple brackets or sliding frictional resistance. The three point bend model did allow for multiple brackets and incorporated sliding frictional resistance. Using these two wire deflection models we were able to determine that the bracket type and archwire diameter have a significant effect on the degree of bracket play, deactivation force and superelastic range during first order tooth movement. This study shows how the ligation mechanism influences the clinician's ability to achieve full first order correction and to maintain light, continuous forces over a wide range of activation. Our results suggest that the clinician can improve

the efficiency of force delivery by considering the degree of correction needed and type of ligation when selecting the appropriate alignment wire.

Conclusions

1. CLBs paired with small diameter superelastic NiTi wires exhibit reduced bracket play and increased superelastic range when compared with SLBs.

2. ASLBs coupled with small diameter superelastic NiTi wires provide reduced bracket play and increased superelastic range when compared with PSLBs.

3. PSLBs require larger diameter wires to provide first order control similar to that of CLBs and ASLBs. If using larger diameter wires, the clinician should consider thermally active wires to reduce the risk of increased binding frictional resistance.

4. When sliding frictional resistance was limited, PSLBs delivered lower deactivation forces when compared to CLBs and ASLBs.

5. Due to their narrow width and passive slide, Damon MX and Carriere brackets exhibited signs of low friction with small diameter wires, but significant binding frictional resistance with a 3mm deflection of large diameter wires.

6. The Damon MX and Carriere PSLBs showed no significant differences for any variable measured.

Figures

Figure 1: Clinical Observation: Derotation with Translation of C_R .

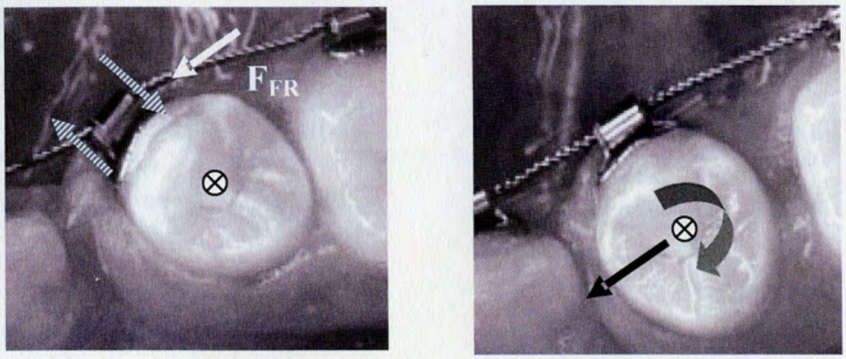


Figure 2: Anatomy of Periodontal Ligament Attachment

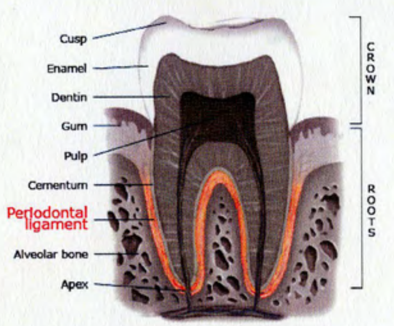


Figure 3: Effect of Continuous Orthodontic Force

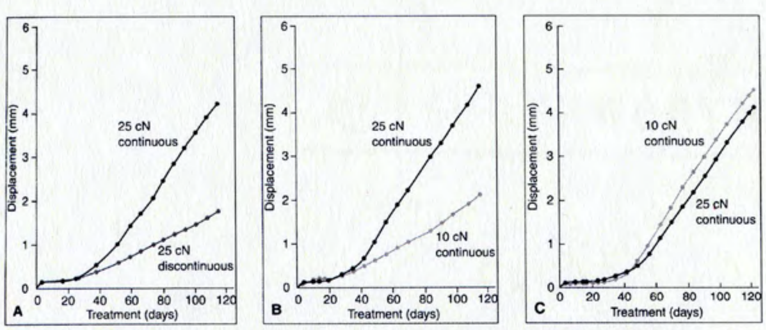


Fig. 2-3 Time displacement curve for premolar tooth movement in a beagle dog experimental model demonstrating that light continuous forces of 25 cN are more effective in tooth movement than discontinuous forces (**A**), and continuous forces of 25 cN produce greater movement than 10 cN in one animal (**B**), while in another animal the two forces produce equal amounts of tooth movement (**C**), demonstrating individual variation and a plateau effect in the latter animal. (Reproduced with permission from van Leeuwen EJ, Maltha JC, Kuijpers-Jagtman AM. Tooth movement with light continuous and discontinuous forces in beagle dogs. Eur J Oral Sci 1999;197:468–474).

Figure 4: Center of Mass and Center of Resistance

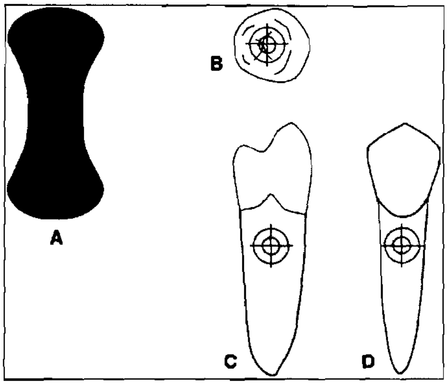


Fig. 1-1 Center of resistance. **A** Center of mass of a free body. **B** Frontal, **C** occlusal, and **D** mesial views of the center of resistance of a single tooth.

Figure 5: Center of Resistance in Relation to Alveolar Bone and Root Length

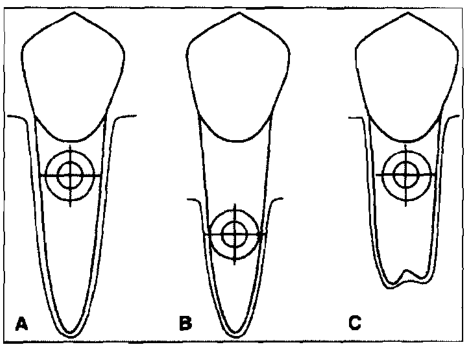


Fig. 1-3 Location of the center of resistance depends on the alveolar bone height and root length. **A** Location of the center of resistance with alveolar bone loss and **B** with a shortened root.

Figure 6: Moment of the Force (M_F)

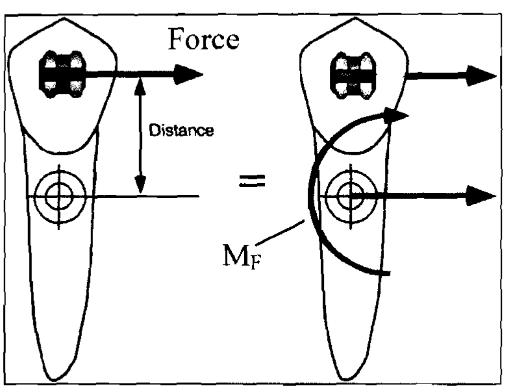


Fig. 1-7 Moment of a force. A force that does not pass through the center of resistance produces a rotational movement as well as linear movement.

Figure 7: Moment of a Couple with Pure Rotation

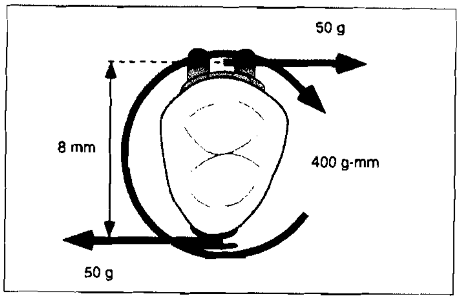


Fig. 1-9 Moment of a couple. A couple produces pure rotation about the center of resistance.

Figure 8: Equivalent Force Systems Diagram

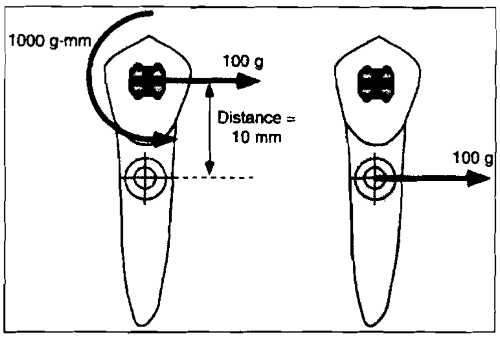


Fig. 1-11 Equivalent force system at a tooth's center of resistance. **A** Force system applied at the bracket. **B** Force system at the center of resistance. The force system at the center of resistance describes the expected tooth movement.

Figure 9: Uncontrolled and Controlled Tipping

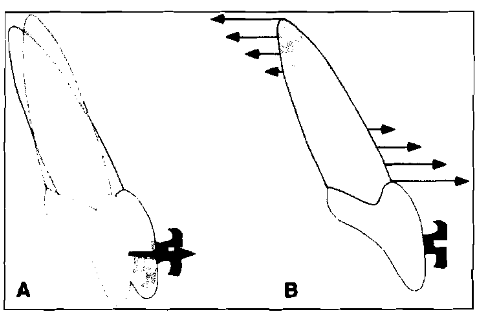


Fig. 1-12 Uncontrolled tipping. **A** Uncontrolled tipping produced by a single force (no applied moment). **B** Stress pattern in the periodontal ligament. Notice the root apex moves in the opposite direction from the movement of the crown.

Figure 10: Bodily Tooth Movement (Translation) and Root Movement (Torque)

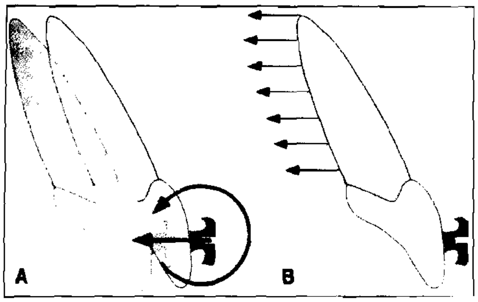


Fig. 1-14 Translation. **A** Translational or bodily tooth movement. **B** Stress pattern in the periodontal ligament with translation. Uniform stresses occur throughout the periodontal ligament.

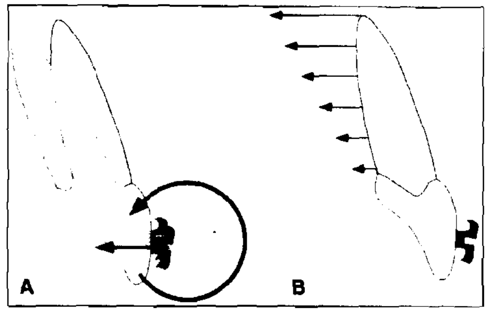


Fig. 1-15 Root movement. **A** Root movement with the center of rotation at the incisal edge. **B** Stress pattern in the periodontal ligament with root movement. The stresses are greatest at the apex.

Figure 11: Pure Rotation about the Center of Resistance

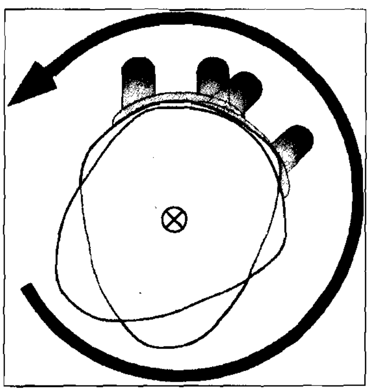


Fig. 1-16 Rotation. Pure rotation occurs around a tooth's center of resistance.

Figure 12: 2^{nd} and 3^{rd} Order Bracket/Archwire Interaction

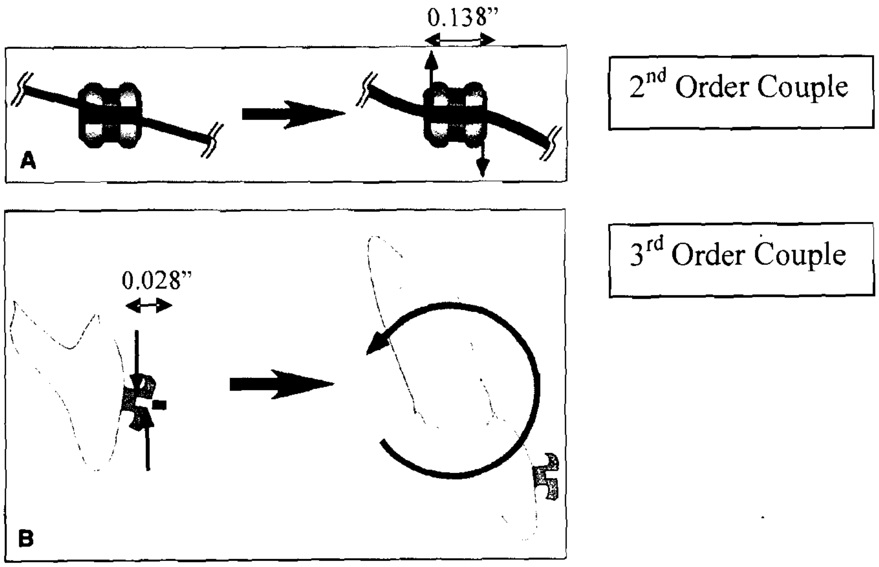


Fig. 1-10 Clinical examples of couples. **A** Engaging a wire in an angulated bracket. **B** Engaging a rectangular (edgewise) wire in a bracket slot.

Figure 13: Stress / Strain Diagram of Archwire Properties

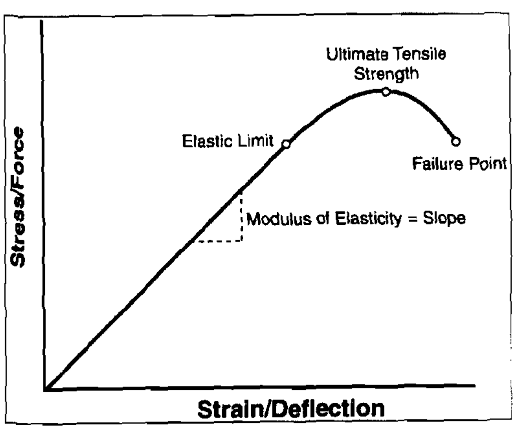


Fig. 1-20 Stress–strain curve for an orthodontic wire. See text for further description of the material characteristics demonstrated.

Figure 14: Comparing Angular and Torsional Deflections

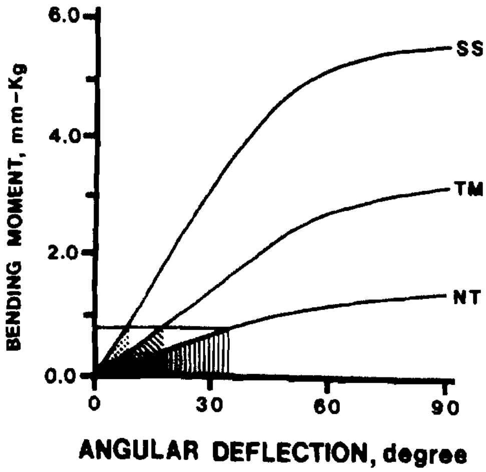


Fig. 1. Stored energy at a fixed bending moment below the proportional limit for 0.019 by 0.025 inch wires of alloys SS, TM, and NT. The stored energy is equal to the shaded area under the curve for each wire. The spring rate is equal to the slope of each curve.

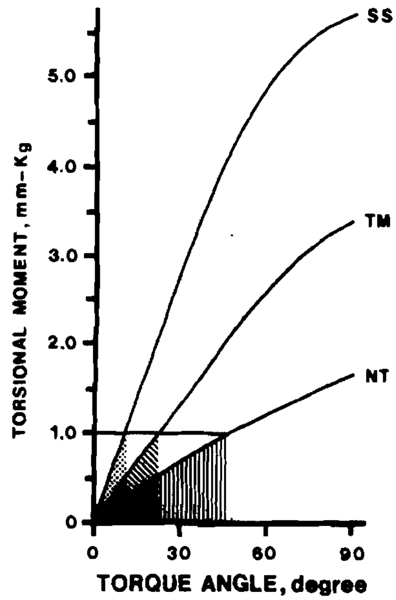


Fig. 2. Stored energy at a fixed torsional moment below the proportional limit for 0.019 by 0.025 inch wires of alloys SS, TM, and NT. The stored energy is equal to the shaded area under the curve for each wire. The spring rate is equal to the slope of each curve.

Figure 15: Moment/Deflection Comparison of Stainless Steel and Nitinol

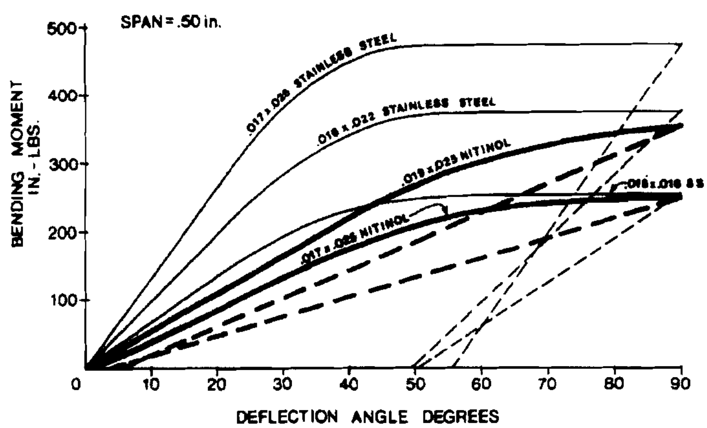


Fig. 2. Comparison of rectangular nitinol and stainless steel arch wires in a bending test, similar to the proposed ADA specifications on orthodontic wire.

Figure 16: Comparison of Chinese NiTi, Nitinol and Stainless Steel

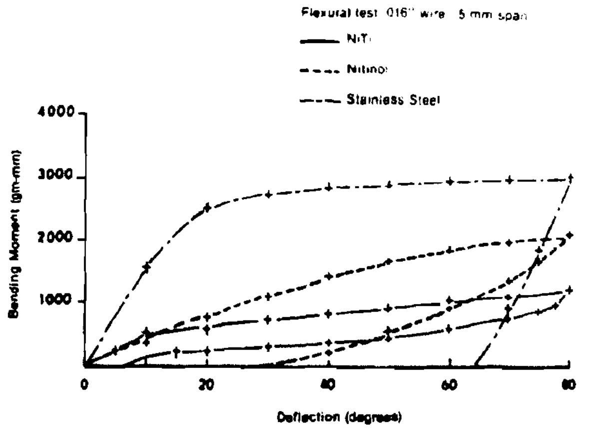


Fig. 2. Bending moment/deflection characteristics of stainless steel, nitinol, and NiTi wires. Both loading (activation) and unloading (deactivation) curves are shown. NiTi wire produces lower moments and forces than nitinol wire.

Figure 17: Nickel Titanium Phase Comparison and Temperature Hysteresis

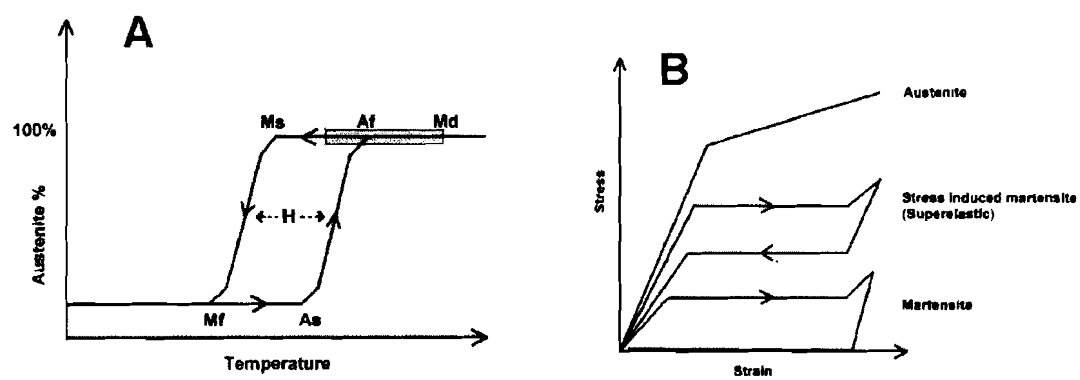


Figure 18: Nickel Titanium Stress Hysteresis

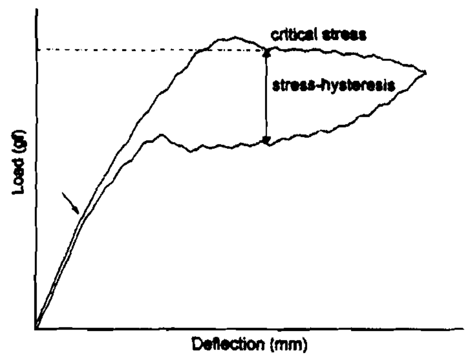


Fig 1. Stress hysteresis, defined as difference between critical stresses in force-deflection curve of Ni-Ti alloy wire, plotted by modified 3-point bending test. Initial martensitic transformation is at origin of nonlinear elasticity (*arrow*).

Figure 19: Comparing Elastomeric Force at Various Time Points

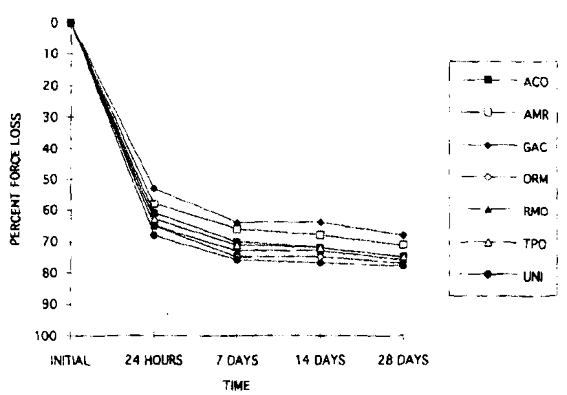


Fig. 7. Percentage of loss of force at initial, 24 hours, 7 days, 14 days, and 28 days.

Figure 20: Pictures of Elastomeric Ligatures Before and After Activation

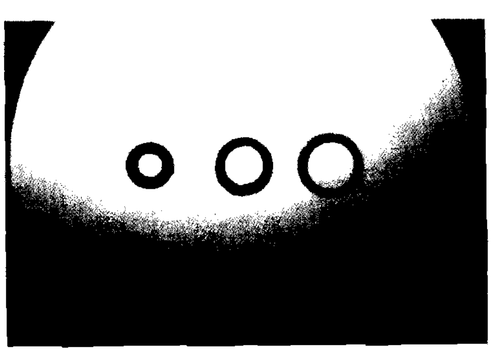


Fig. 4. From left to right, ligatures before testing, after storage (stretched) for 28 days at room temperature and humidity, and after storage (stretched) for 28 days in simulated oral environment. (Magnification ×4.)

Figure 21: Passive Ligation Mechanism

Figure 22: Interactive Ligation Mechanism

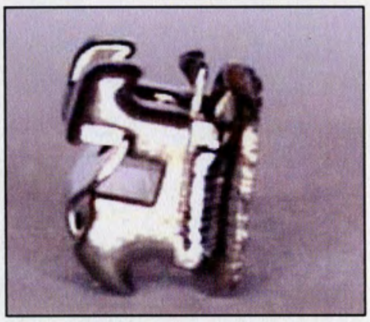
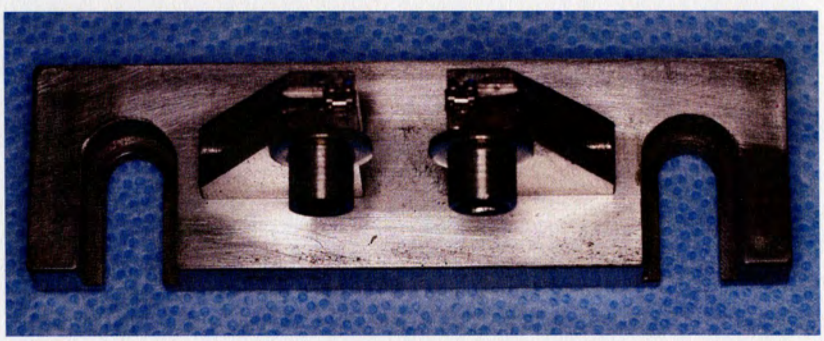


Figure 23: Custom Testing Fixture with Mounted Brackets



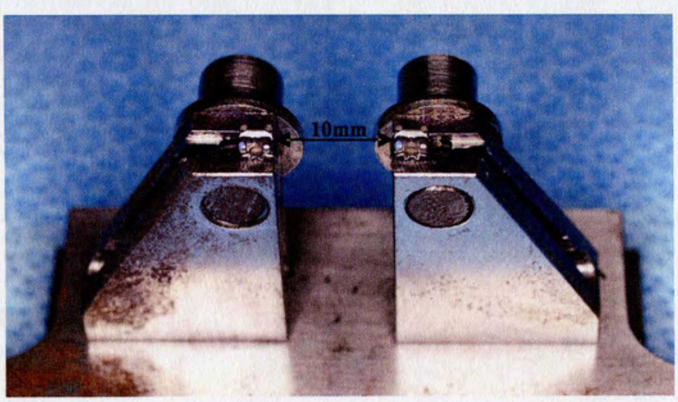


Figure 24: Tinius Olsen H1K-S Mechanical Testing Instrument

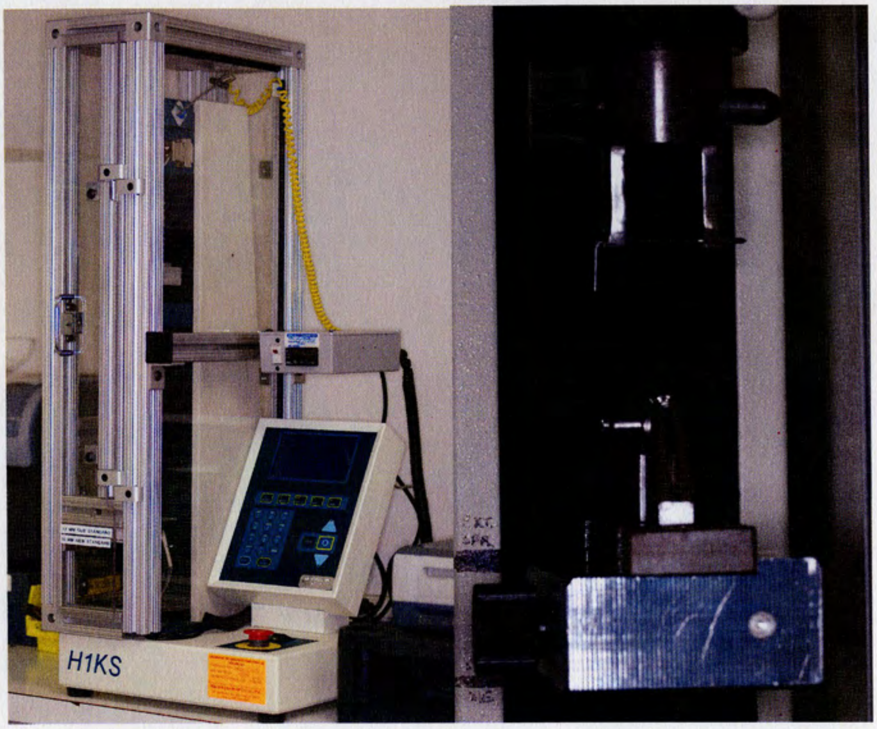


Figure 25: Example Load-Deflection Curve

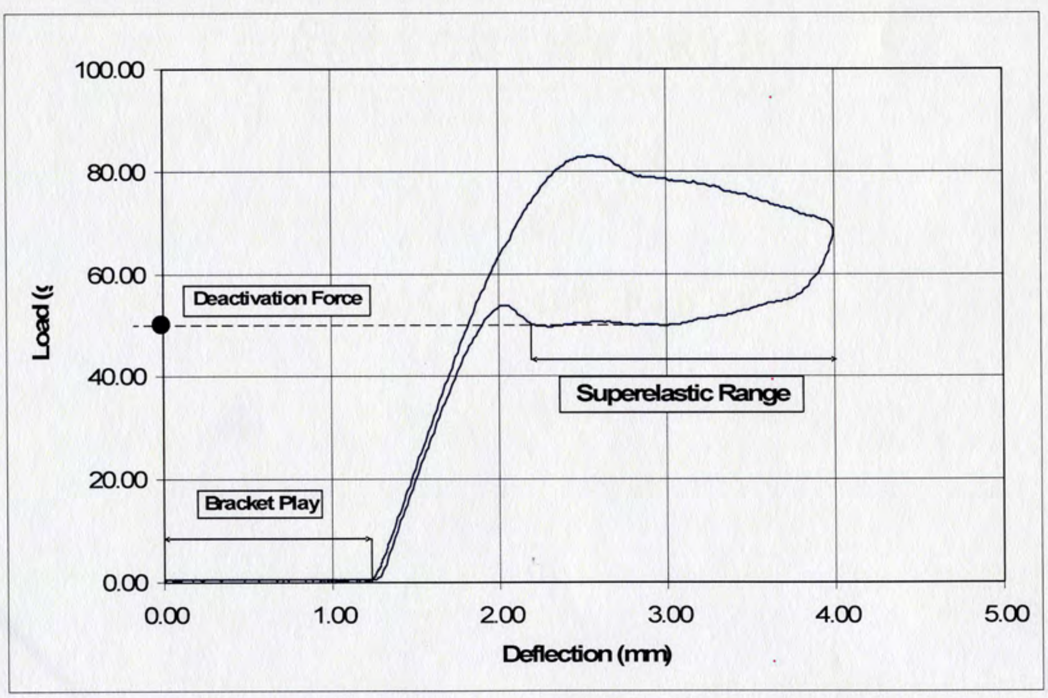


Figure 26A: Load-Deflection Curves for 0.014" SE NiTi Cantilever Test

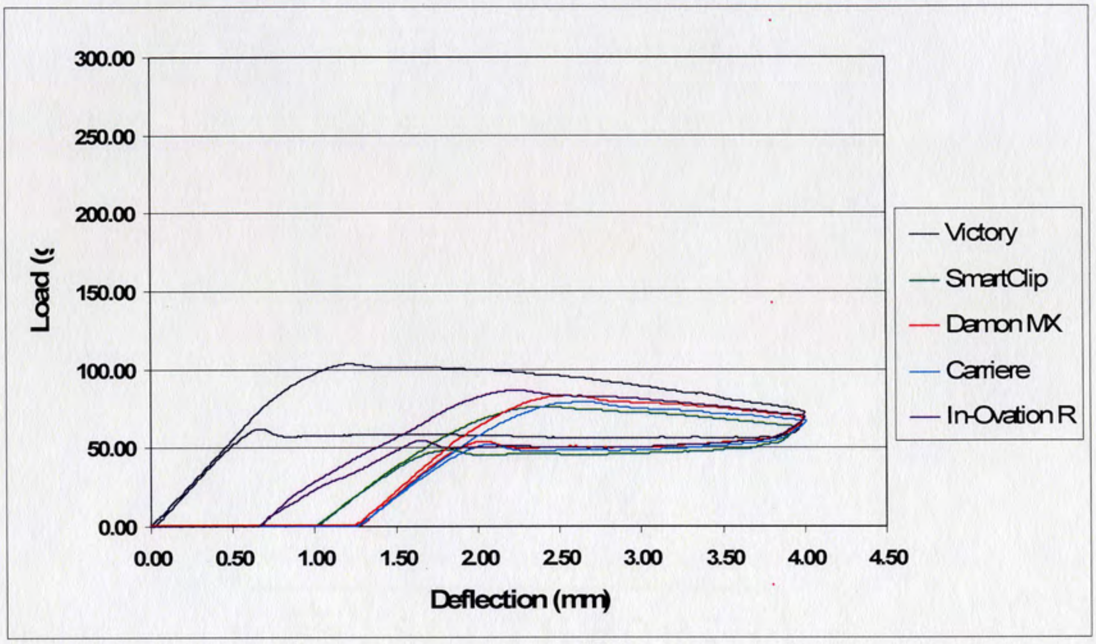


Figure 26B: Load-Deflection Curves for 0.016" SE NiTi Cantilever Test

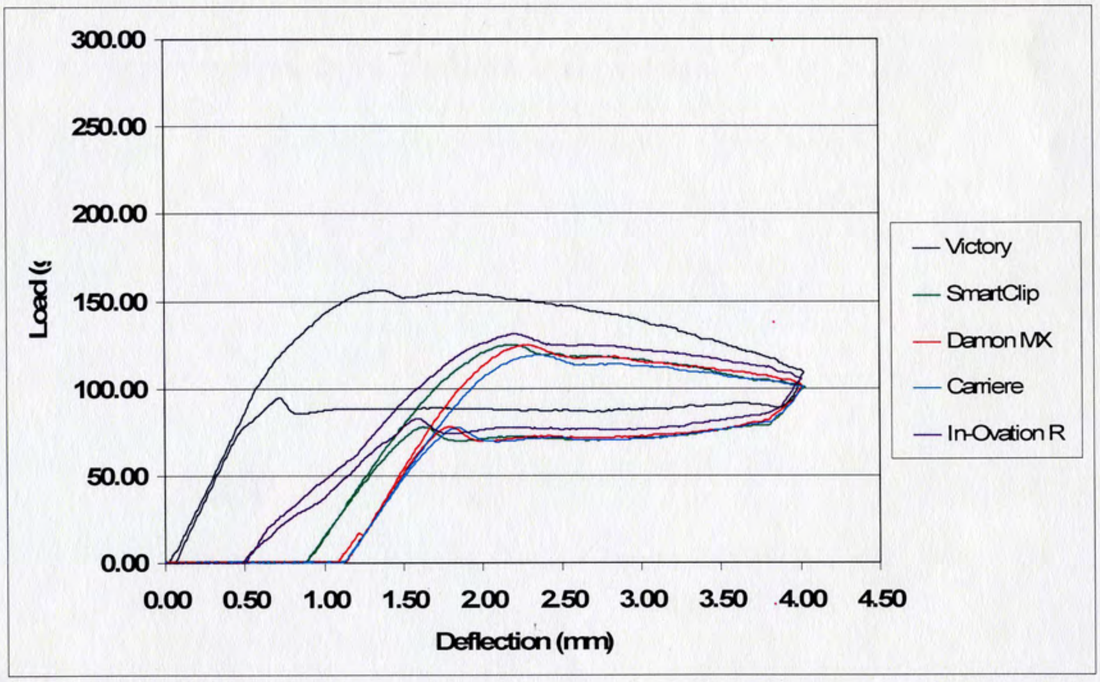


Figure 26C: Load-Deflection Curves for 0.018" SE NiTi Cantilever Test

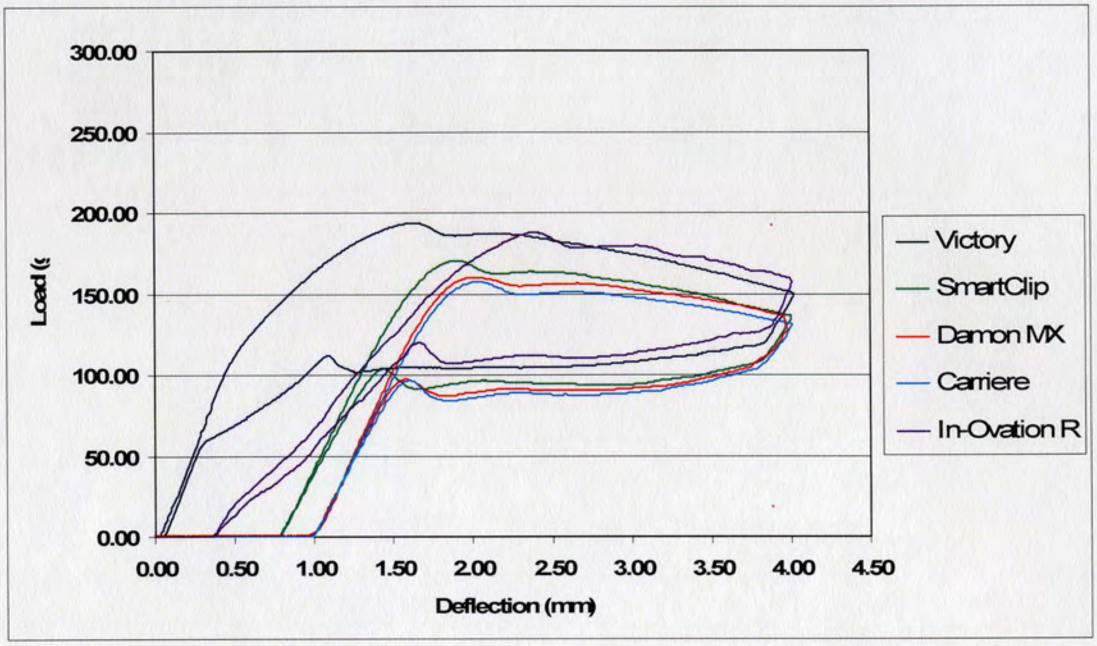


Figure 26D: Load-Deflection Curves for 0.020" SE NiTi Cantilever Test

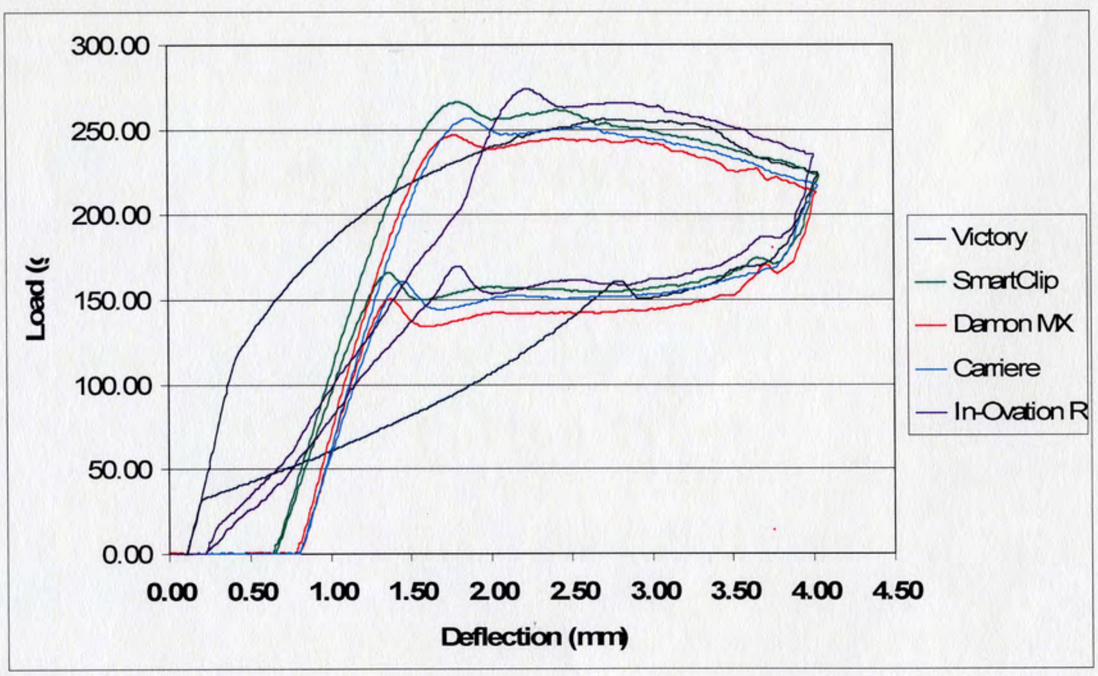


Figure 27: Mean Degrees of Bracket Play with Cantilever Test (* $p \leq 0.01$)

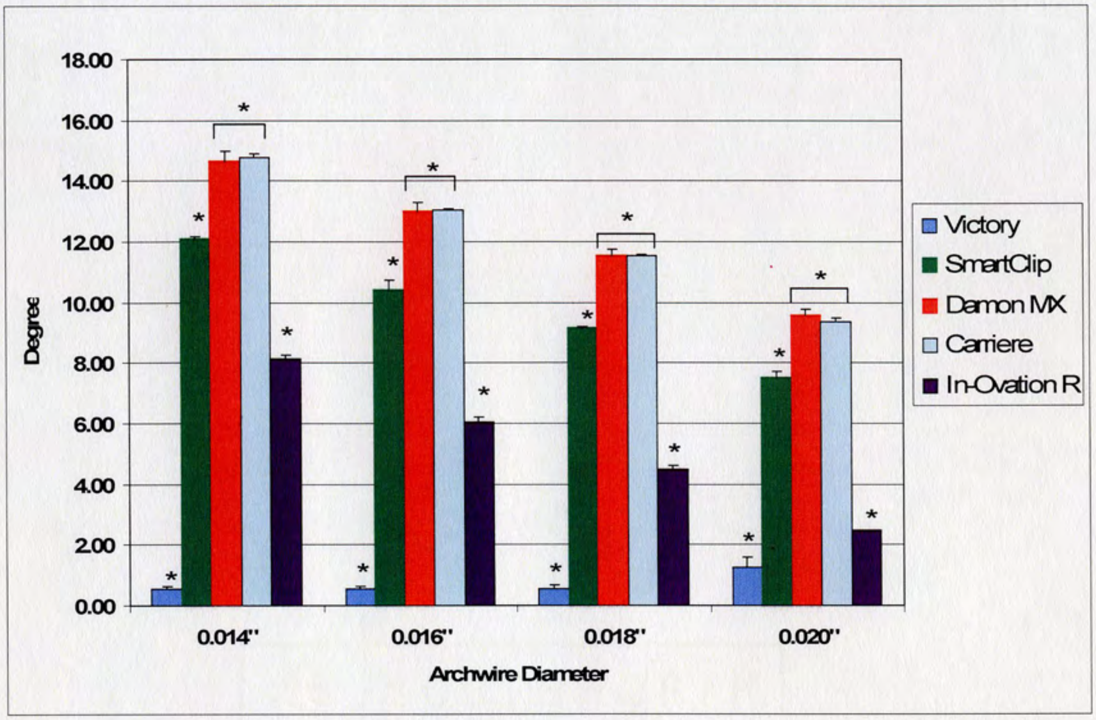


Figure 28: Mean Deactivation Force with 1mm Cantilever Deflection

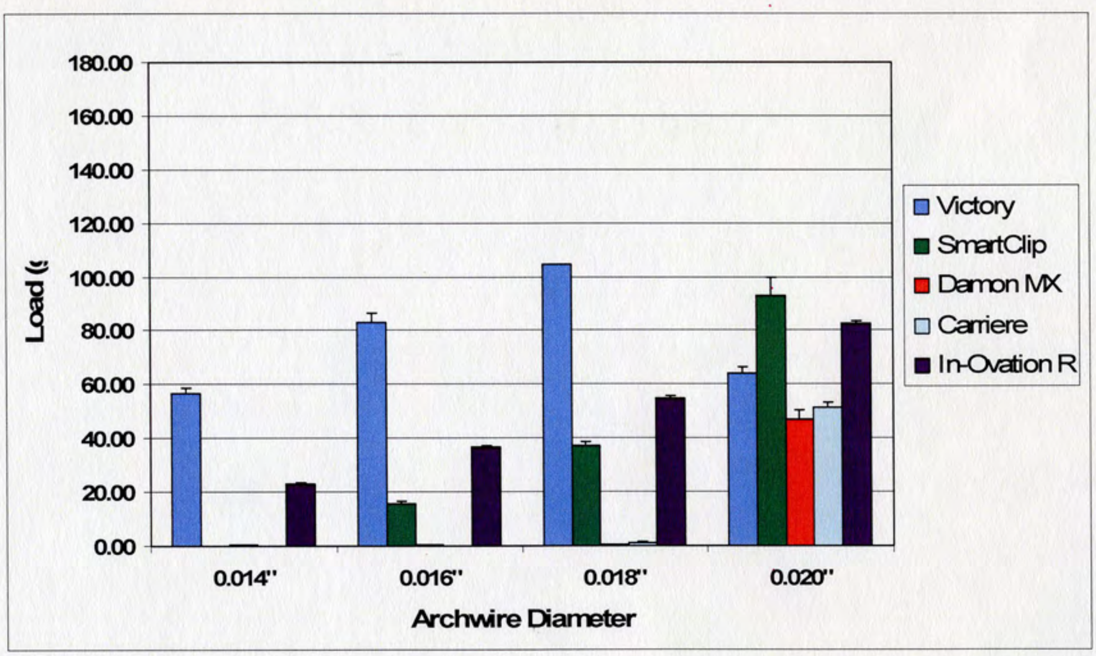


Figure 29: Mean Deactivation Force with 2mm Cantilever Deflection

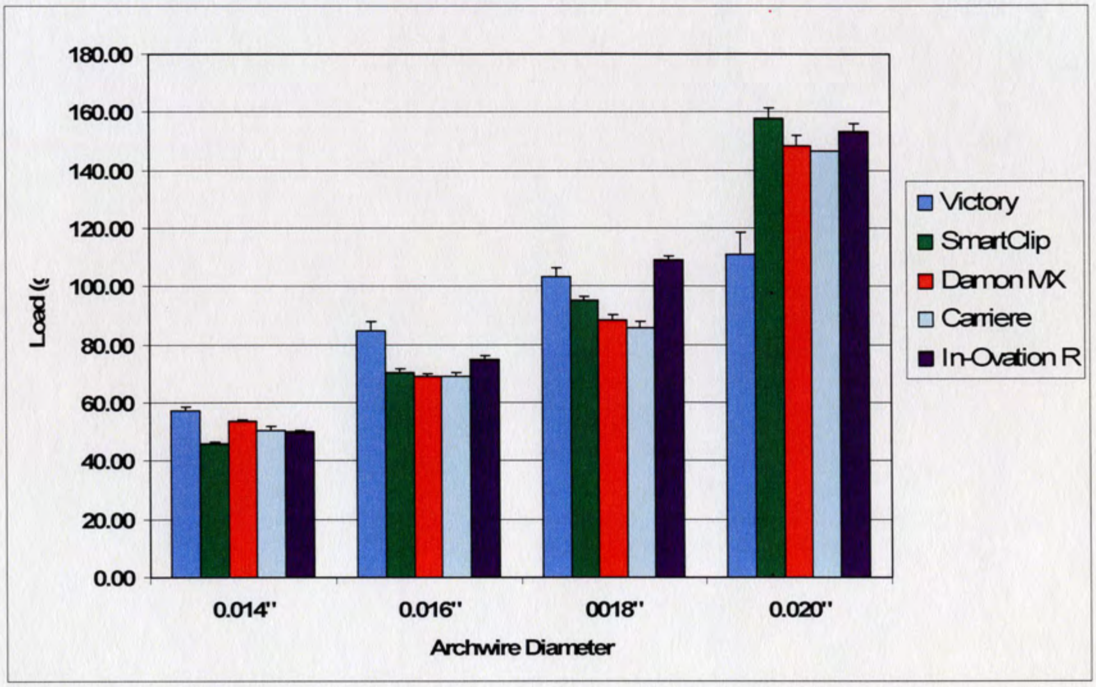


Figure 30: Mean Deactivation Force with 3mm Cantilever Deflection

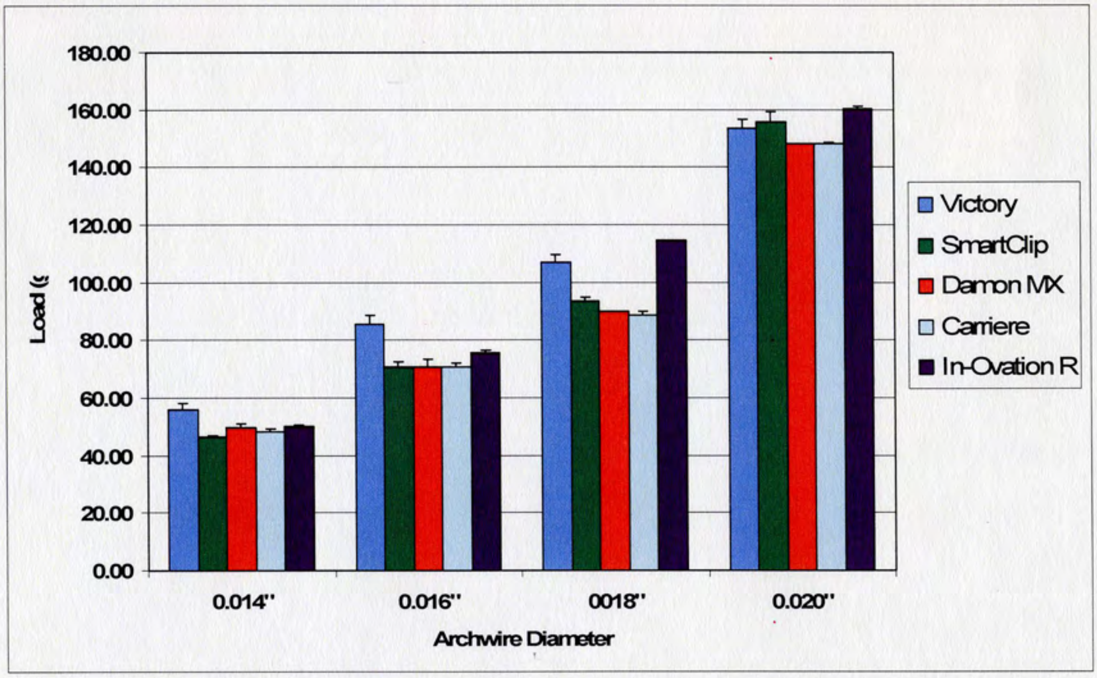


Figure 31: Mean Superelastic Range with Cantilever Test (* $p \leq 0.01$)

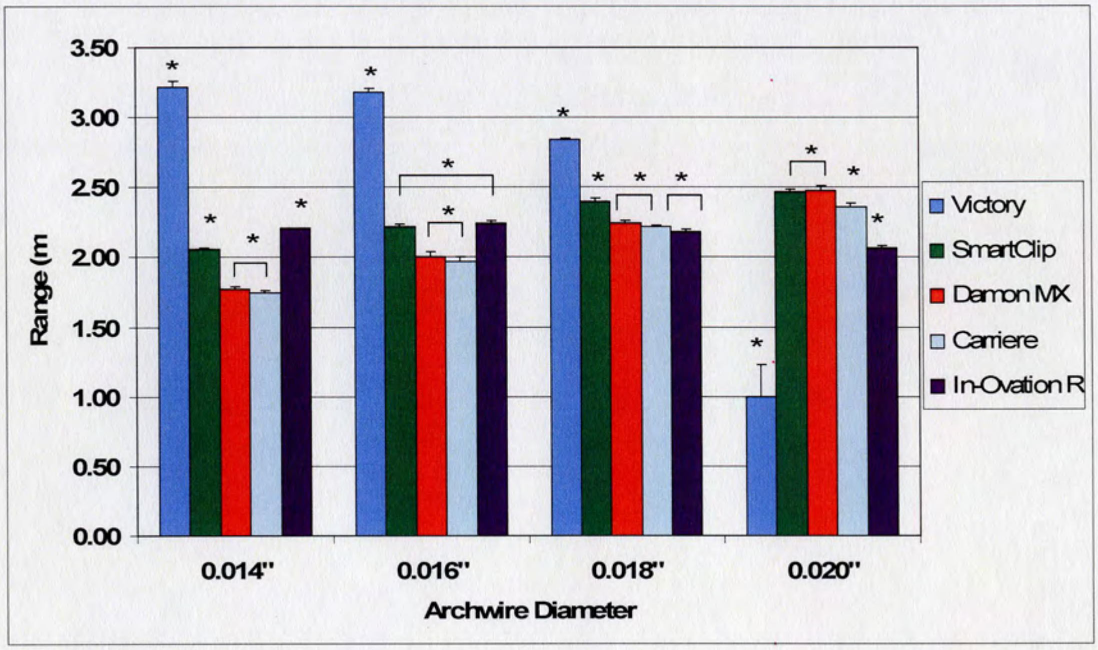


Figure 32A: Load-Deflection Curves for 0.014" SE NiTi Three Point Bend Test

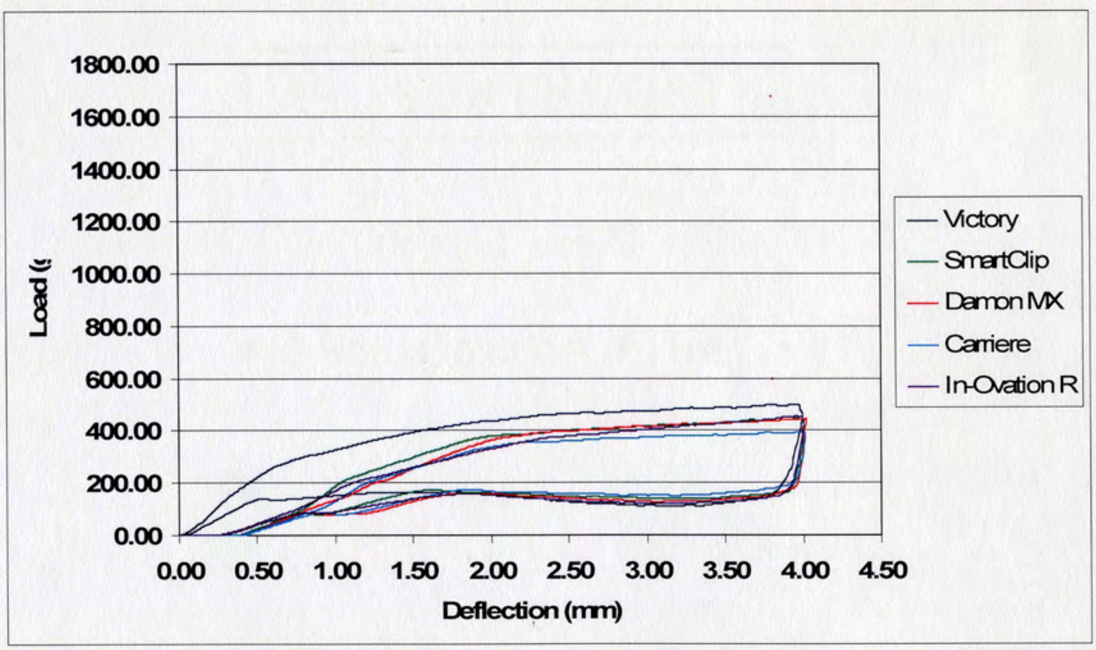


Figure 32B: Load-Deflection Curves for 0.016" SE NiTi Three Point Bend Test

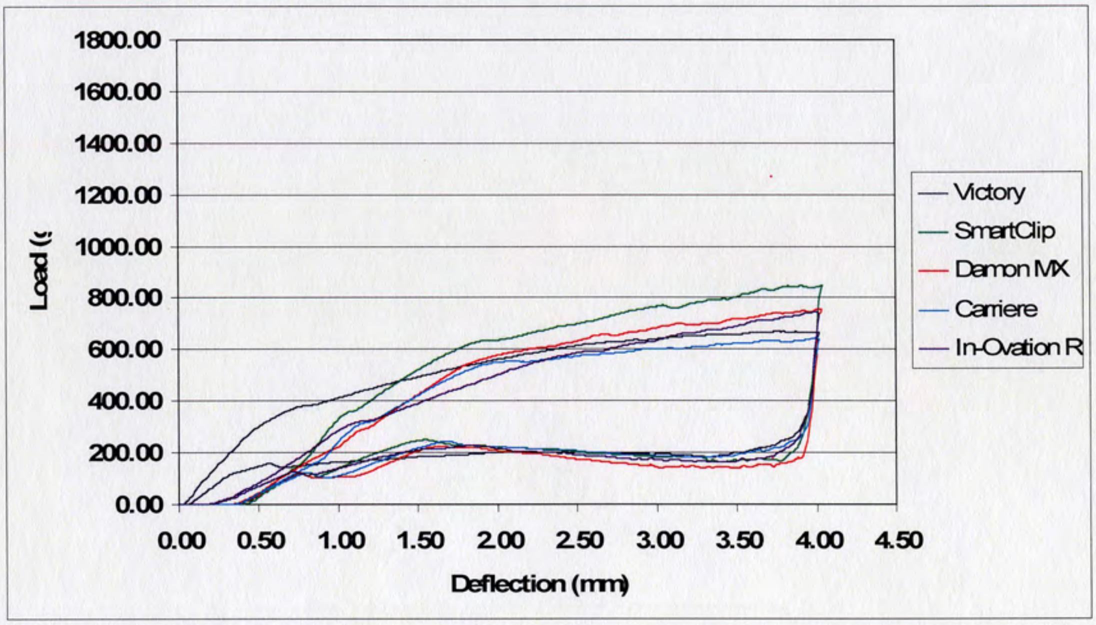


Figure 32C: Load-Deflection Curves for 0.018" SE NiTi Three Point Bend Test

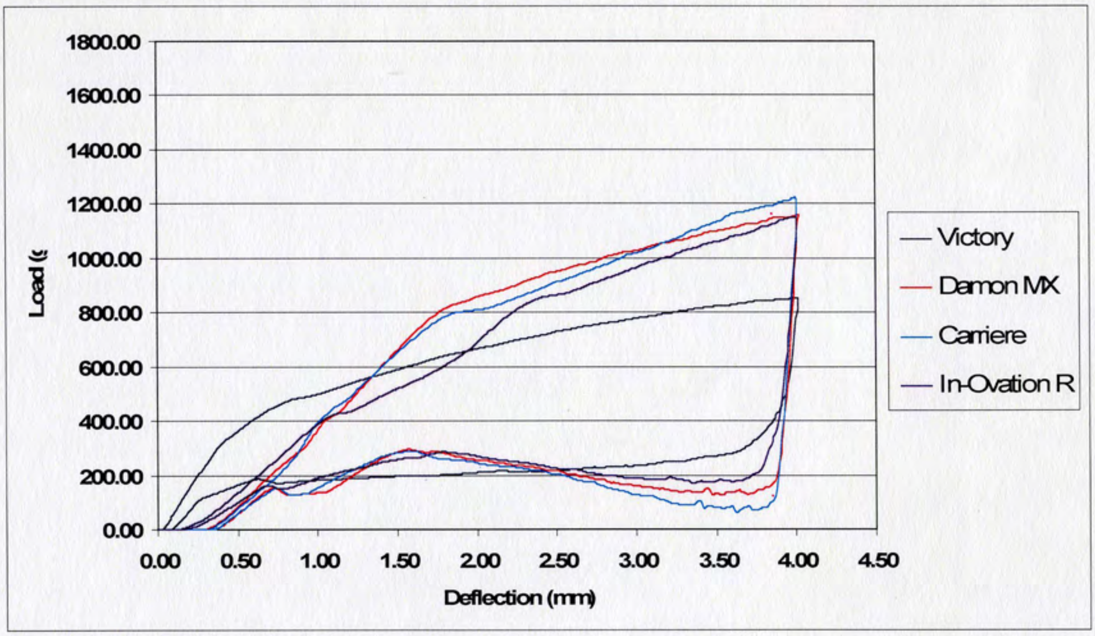


Figure 32D: Load-Deflection Curves for 0.020" SE NiTi Three Point Bend Test

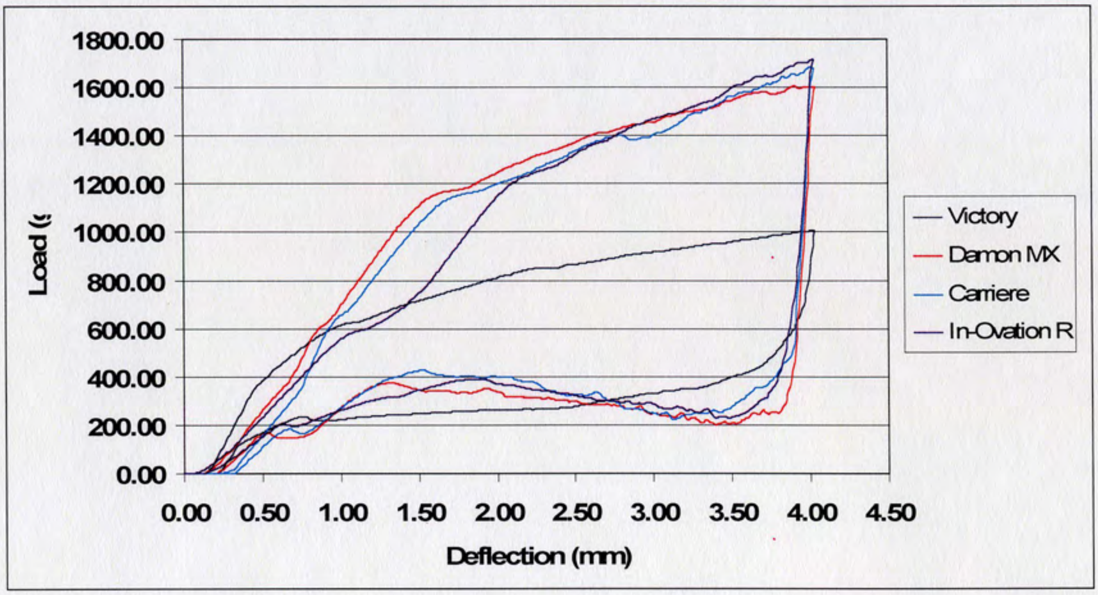


Figure 33: Mean Deactivation Forces at 1mm of Three Point Deflection

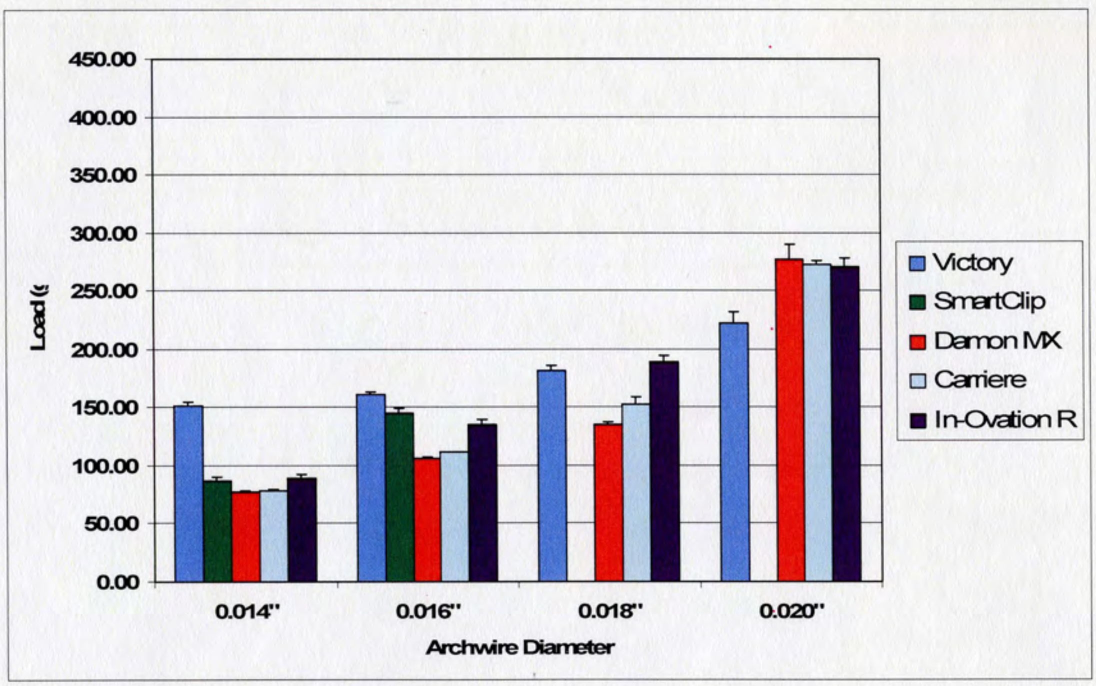


Figure 34: Mean Deactivation Forces at 2mm of Three Point Deflection

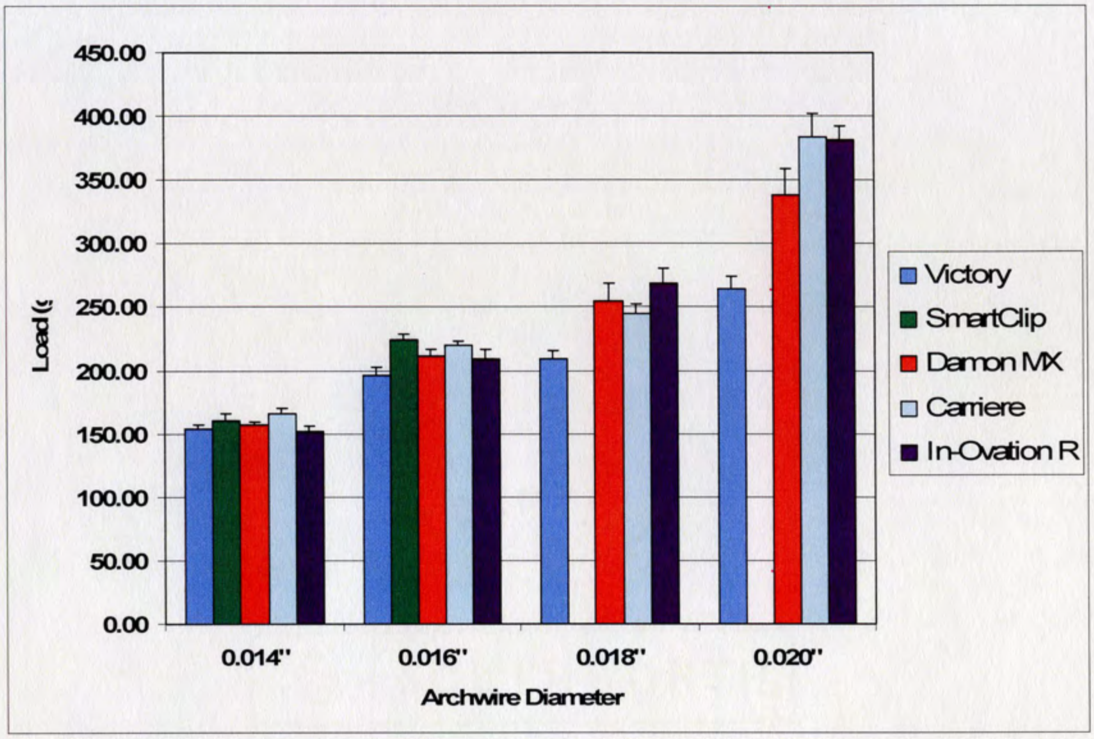
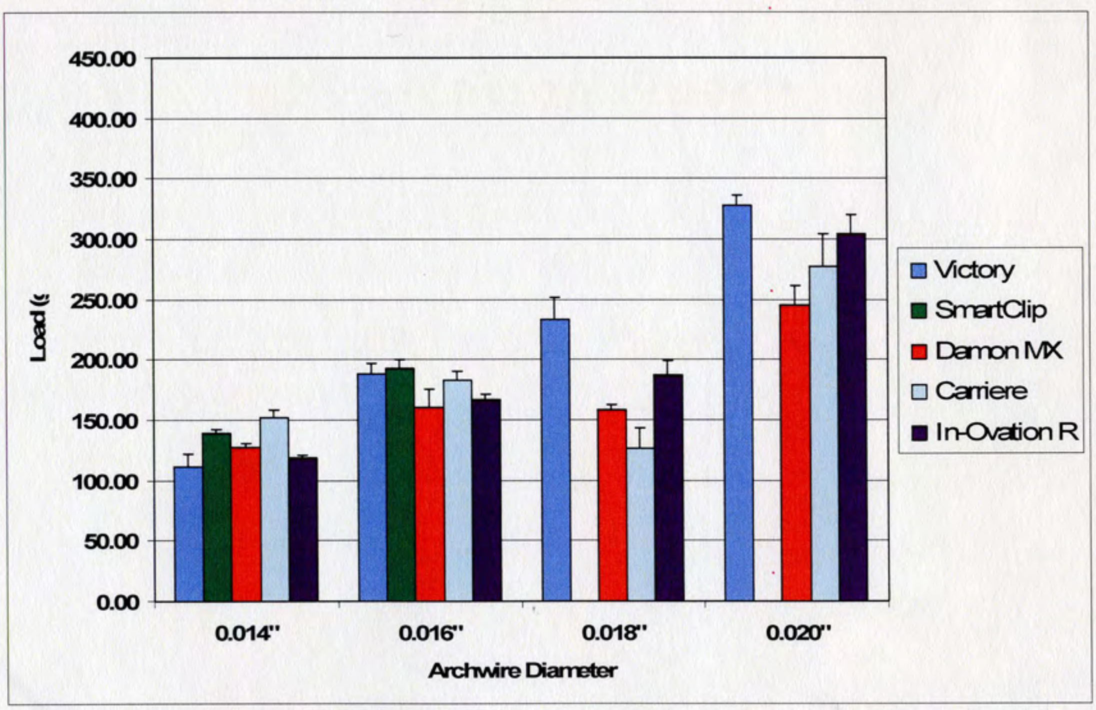


Figure 35: Mean Deactivation Forces at 3mm of Three Point Deflection



<u>Tables</u>

Table I: Description of Brackets Tested.

Bracket Type	Ligation Mechanism	Slot Width	Effective Width*	Material	Manufacturer
Victory™	Conventional Elastomeric	2.98mm	3.73mm	Stainless Steel	3M Unitek
SmartClip™	Passive Clips	4.15mm	3.80mm	Stainless Steel NiTi Clips	3M Unitek
Damon MX™	Passive Slide	2.67mm	2.39mm	Stainless Steel	Ormco
Carriere™	Passive Slide	3.35mm	2.84mm	Stainless Steel	Class One
In-Ovation R™	Interactive Clip	2.97mm	2.97mm	Stainless Steel Elgiloy Door	GAC Intl.
* Distance between the contact points formed during a first-order couple. Varies with the width of the slot base and the ligation mechanism.					

Table II: Description of Wires Tested.

Material	TTR	Diameters				Manufacturer
SE NiTi	55° F	0.014"	0.016"	0.018"	0.020"	Ultimate Wireforms Inc., Bristol, CT

Table III: Degrees of Bracket Play with 0.014", 0.016", 0.018" and 0.020" SE NiTi

Bracket Play - Mean degrees of archwire deflection required to produce 5g of loading force.						
		Victory	**SmartClip**	**Damon MX**	**Carriere**	**In-Ovation R**
0.014"						
	Mean	0.57	12.08	14.66	14.79	8.13
	S.D.	0.004	0.010	0.029	0.010	0.013
	Post-hoc*	A	B	C	C	D
0.016"						
	Mean	0.56	10.39	12.99	13.06	6.03
	S.D.	0.007	0.034	0.031	0.004	0.016
	Post-hoc*	A	B	C	C	D
0.018"						
	Mean	0.55	9.13	11.56	11.52	4.51
	S.D.	0.010	0.006	0.016	0.005	0.011
	Post-hoc*	A	B	C	C	D
0.020"						
	Mean	1.25	7.51	9.56	9.36	2.49
	S.D.	0.031	0.024	0.019	0.010	0.028
	Post-hoc*	A	B	C	C	D
* Means with same letter are not significantly different at the 0.01 level of significance.						

Table IV - A: Cantilever Deactivation Force at 1mm, 2mm and 3mm Deflection

0.014" SE NiTi Buccal Cantilever Deflection					
	Victory	**SmartClip**	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	56.67	0.16	0.38	0.135	22.74
	55.42	0.065	0.64	0.083	23.25
	58.8	0.34	0.17	0.169	22.31
Mean	56.96	0.15	0.40	0.13	22.77
Std Dev	1.71	0.20	0.24	0.04	0.47
Post Hoc*	A	B	B	B	C
2mm	57.09	46.57	52.65	48.91	50.48
	55.40	45.54	54.05	51.23	50.51
	58.74	45.72	53.95	51.11	49.80
Mean	57.08	45.94	53.55	50.42	50.26
Std Dev	1.67	0.55	0.78	1.31	0.40
Post Hoc*	A	B	C	D	D
3mm	55.91	46.92	48.88	47.76	50.74
	53.71	45.97	50.48	48.93	49.84
	57.98	46.38	50.25	48.95	50.14
Mean	55.87	46.42	49.87	48.55	50.24
Std Dev	2.14	0.48	0.87	0.68	0.46
Post Hoc*	A	B	C	C	C
* Means with same letter are not significantly different at the 0.01 level of significance					

Table IV - B: Cantilever Deactivation Force at 1mm, 2mm and 3mm Deflection

0.016" SE NiTi Buccal Cantilever Deflection					
	Victory	**SmartClip**	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	87.19	15.11	0.51	0.17	36.99
	80.57	16.92	0.68	0.17	36.49
	82.43	15.25	0.34	0.10	37.05
Mean	83.40	15.76	0.51	0.15	36.84
Std Dev	3.41	1.01	0.17	0.04	0.31
Post-hoc*	A	B	C	C	D
2mm	88.16	68.67	67.98	69.33	73.47
	81.69	71.67	69.87	67.79	75.04
	84.36	70.84	69.42	70.23	76.30
Mean	84.74	70.39	69.09	69.12	74.94
Std Dev	3.25	1.55	0.99	1.23	1.42
Post-hoc*	A	B	B	B	C
3mm	87.87	68.84	68.27	71.77	74.36
	82.39	72.06	72.61	69.50	75.80
	86.68	71.38	71.89	70.92	76.67
Mean	85.65	70.76	70.92	70.73	75.61
Std Dev	2.88	1.70	2.33	1.15	1.17
Post-hoc*	A	B	B/C	B	C
* Means with same letter are not significantly different at the 0.01 level of significance.					

Table IV - C: Cantilever Deactivation Force at 1mm, 2mm and 3mm Deflection

0.018" SE NiTi Buccal Cantilever Deflection					
	Victory	**SmartClip**	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	107.16	36.59	0.28	0.44	55.74
	103.12	38.96	0.41	1.52	55.02
	103.47	35.33	0.59	0.55	53.74
Mean	104.58	36.96	0.43	0.84	54.83
Std Dev	2.24	1.84	0.16	0.59	1.01
Post Hoc*	A	B	C	C	D
2mm	105.55	94.10	86.95	83.33	108.51
	100.26	96.54	87.79	85.25	110.56
	104.73	95.53	90.25	88.04	108.05
Mean	103.51	95.39	88.33	85.54	109.04
Std Dev	2.85	1.23	1.71	2.37	1.34
Post Hoc*	A	B	C	C	A
3mm	109.08	92.91	88.30	88.13	114.47
	104.20	94.90	91.13	88.34	114.32
	107.75	92.36	91.21	90.01	114.55
Mean	107.01	93.39	90.21	88.83	114.45
Std Dev	2.52	1.34	1.66	1.03	0.12
Post Hoc*	A	B	B/C	C	D
* Means with same letter are not significantly different at the 0.01 level of significance.					

Table IV - D: Cantilever Deactivation Force at 1mm, 2mm and 3mm Deflection

0.020" SE NiTi Buccal Cantilever Deflection					
	Victory	**SmartClip**	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	66.77	86.12	46.48	53.24	83.04
	66.98	99.60	50.66	49.00	81.24
	58.96	92.87	43.46	51.79	83.60
Mean	64.24	92.86	46.87	51.34	82.63
Std Dev	4.57	6.74	3.62	2.16	1.23
Post-hoc*	A	B	C	C	D
2mm	116.33	154.26	144.46	146.50	150.60
	114.15	161.70	151.18	146.63	152.96
	101.94	157.49	150.22	146.62	156.42
Mean	110.81	157.82	148.62	146.58	153.33
Std Dev	7.76	3.73	3.63	0.07	2.93
Post-hoc*	A	B	B/C	C	B
3mm	152.62	151.77	145.31	147.92	159.29
	151.23	158.89	149.90	148.88	159.99
	157.06	156.70	149.01	147.35	161.09
Mean	153.64	155.79	148.07	148.05	160.12
Std Dev	3.05	3.65	2.43	0.77	0.91
Post-hoc*	A/B	A/B	A	A	B
* Means with same letter are not significantly different at the 0.01 level of significance.					

Table V: Mean Superelastic Range with 4mm Buccal Archwire Deflection

		Victory	SmartClip	Damon MX	Carriere	In-Ovation R
0.014"	Mean	3.22	2.05	1.77	1.75	2.21
	S.D.	0.04	0.02	0.02	0.02	0.01
	Post-hoc*	A	B	C	C	D
0.016"	Mean	3.18	2.21	2.01	1.97	2.25
	S.D.	0.03	0.03	0.03	0.04	0.02
	Post-hoc*	A	B	C	C	B
0.018"	Mean	2.84	2.40	2.24	2.22	2.18
	S.D.	0.01	0.03	0.02	0.01	0.02
	Post-hoc*	A	B	C	C/D	D
0.020"	Mean	0.99	2.46	2.48	2.36	2.06
	S.D.	0.24	0.02	0.04	0.03	0.02
	Post-hoc*	A	B	B	C	D

* Means with same letter are not significantly different at the 0.01 level of significance.

Table VI - A: Three Point Deactivation Force at 1mm, 2mm and 3mm Deflection

0.014" SE NiTi Buccal Three Point Bend Deflection					
	Victory	**SmartClip**	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	151.26	90.80	78.43	77.98	93.19
	154.46	86.40	78.02	79.98	86.64
	149.74	83.77	76.50	78.06	87.92
AVG	151.82	86.99	77.65	78.67	89.25
STDV	2.41	3.55	1.02	1.13	3.47
Post-Hoc*	A	B	C	C	B
2mm	152.76	162.82	159.35	169.78	156.33
	157.85	155.11	154.97	167.06	151.81
	152.12	164.74	156.01	162.17	148.3
AVG	154.24	160.89	156.78	166.34	152.15
STDV	3.14	5.10	2.29	3.86	4.03
Post-Hoc*	A	A/B	A	B	A
3mm	99.93	141.84	124.71	158.38	120.31
	119.84	140.26	131.23	152.27	119.41
	116.01	135.11	127.44	146.15	116.84
AVG	111.93	139.07	127.79	152.27	118.85
STDV	10.56	3.52	3.27	6.12	1.80
Post-Hoc*	A	B	A	B	A
* Means with same letter are not significantly different at the 0.01 level of significance.					

Table VI - B: Three Point Deactivation Force at 1mm, 2mm and 3mm Deflection

0.016" SE NiTi Buccal Three Point Bend Deflection					
	Victory	**SmartClip**	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	162.01	150.36	107.52	110.61	131.27
	162.86	142.44	105.94	112.33	138.06
	158.30	142.82	105.61	111.03	137.86
AVG	161.06	145.21	106.36	111.32	135.73
STDV	2.42	4.47	1.02	0.90	3.86
Post-Hoc*	A	B	C	D	B
2mm	199.43	225.40	217.70	219.90	200.70
	200.80	228.60	208.50	217.60	215.60
	187.90	219.30	206.80	223.70	210.80
AVG	196.04	224.43	211.00	220.40	209.03
STDV	7.09	4.72	5.86	3.08	7.61
Post-Hoc*	A	B	A/B	B	A
3mm	192.48	201.60	175.57	174.07	168.94
	194.16	188.70	161.97	187.02	170.46
	178.22	189.07	145.08	188.30	162.00
AVG	188.29	193.12	160.87	183.13	167.13
STDV	8.76	7.34	15.27	7.87	4.51
Post-Hoc*	A	A	A	A	A

* Means with same letter are not significantly different at the 0.01 level of significance.

Table VI - C: Three Point Deactivation Force at 1mm, 2mm and 3mm Deflection

0.018" SE NiTi Buccal Three Point Bend Deflection				
	Victory	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	178.75	137.62	155.35	188.38
	186.60	133.77	144.54	183.09
	179.92	133.50	156.11	194.19
AVG	181.76	134.96	152.00	188.55
STDV	4.24	2.30	6.47	5.55
Post-Hoc*	A	B	C	A
2mm	205.30	269.10	252.40	280.30
	216.10	254.20	237.30	256.50
	207.50	242.30	244.50	269.60
AVG	209.63	255.20	244.73	268.80
STDV	5.71	13.43	7.55	11.92
Post-Hoc*	A	B	B	B
3mm	224.80	164.06	141.71	200.40
	255.00	156.59	108.36	178.59
	222.50	156.56	130.76	183.92
AVG	234.10	159.07	126.94	187.64
STDV	18.14	4.32	17.00	11.37
Post-Hoc*	A	B	B	C
* Means with same letter are not significantly different at the 0.01 level of significance.				

Table VI - D: Three Point Deactivation Force at 1mm, 2mm and 3mm Deflection

0.020" SE NiTi Buccal Three Point Bend Deflection				
	Victory	**Damon MX**	**Carriere**	**In-Ovation R**
1mm	211.70	291.30	272.30	277.30
	231.70	273.40	276.30	272.20
	223.10	266.40	270.50	261.40
AVG	222.17	277.03	273.03	270.30
STDV	10.03	12.84	2.97	8.12
Post-Hoc*	A	B	B	B
2mm	256.30	359.80	385.80	393.20
	275.30	318.10	365.60	381.20
	263.10	337.70	401.30	370.70
AVG	264.90	338.53	384.23	381.70
STDV	9.63	20.86	17.90	11.26
Post-Hoc*	A	B	B	B
3mm	317.30	259.70	299.10	322.10
	334.50	227.70	285.60	300.60
	330.90	249.40	249.30	289.40
AVG	327.57	245.60	278.00	304.03
STDV	9.07	16.33	25.76	16.62
Post-Hoc*	A	B	A/B	A
* Means with same letter are not significantly different at the 0.01 level of significance.				

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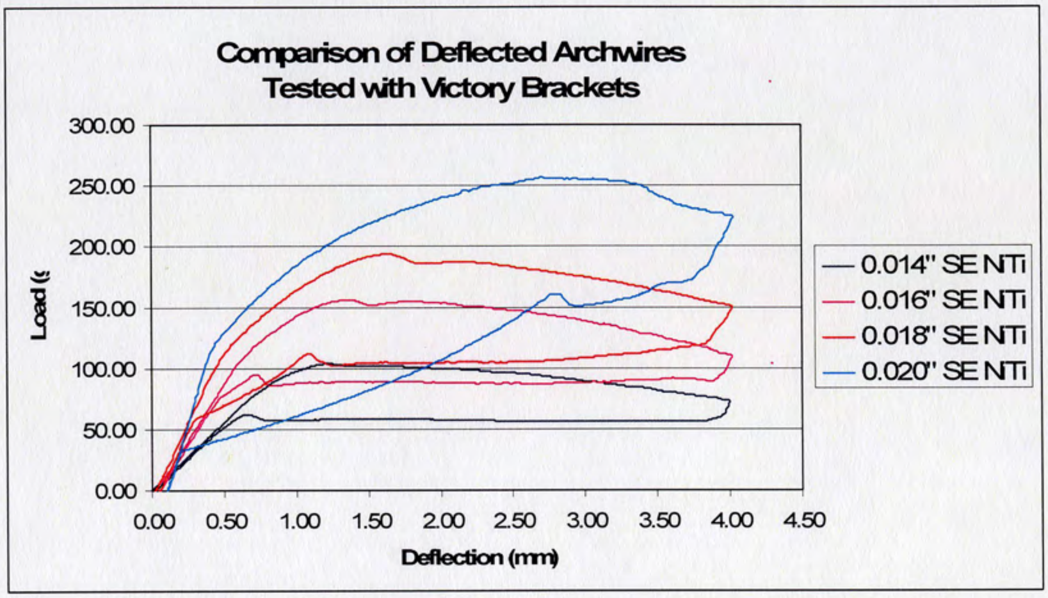
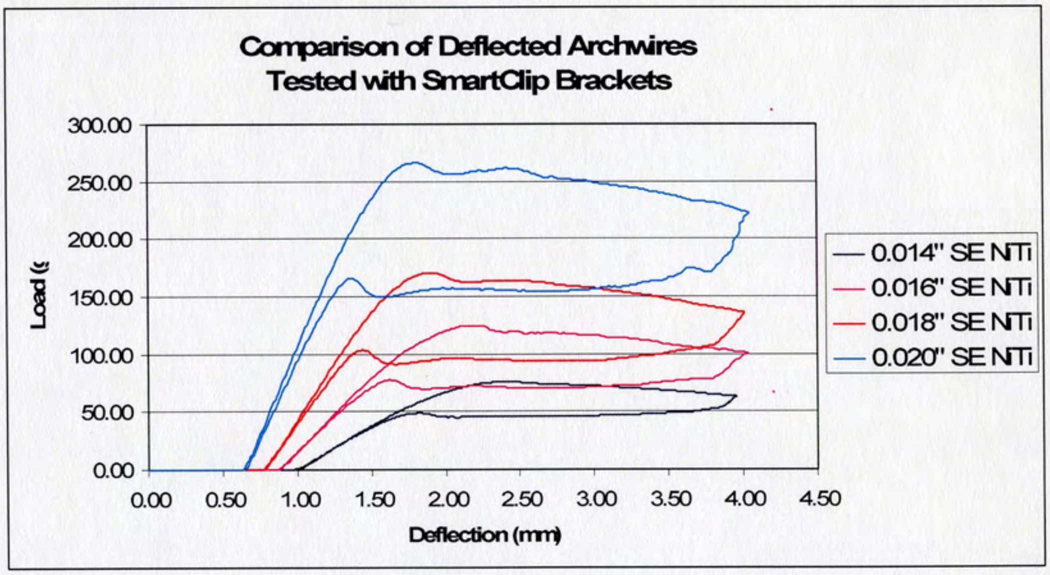
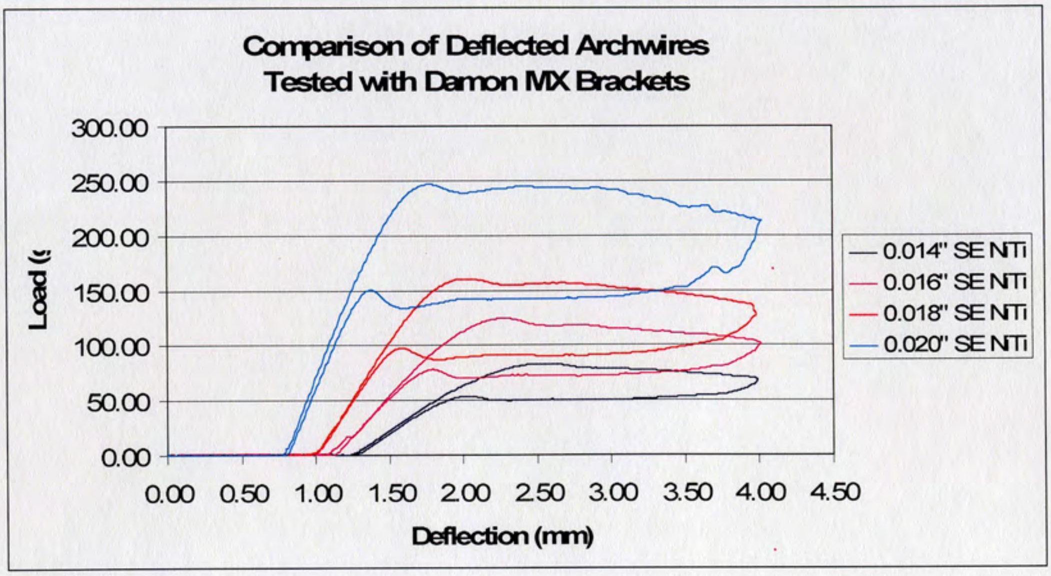
Appendix

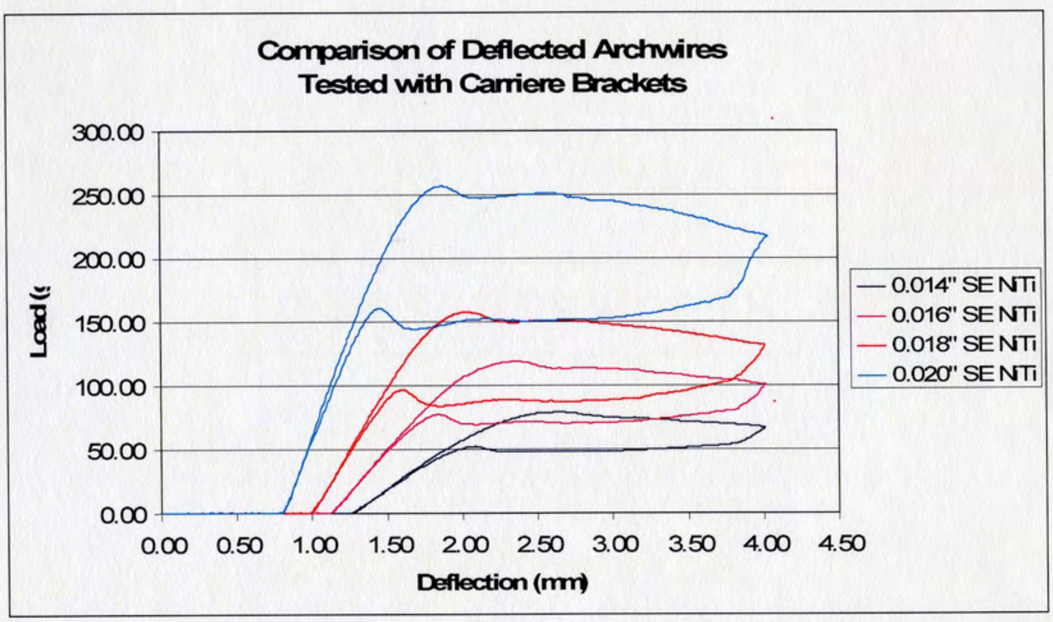
Cantilever Graphs



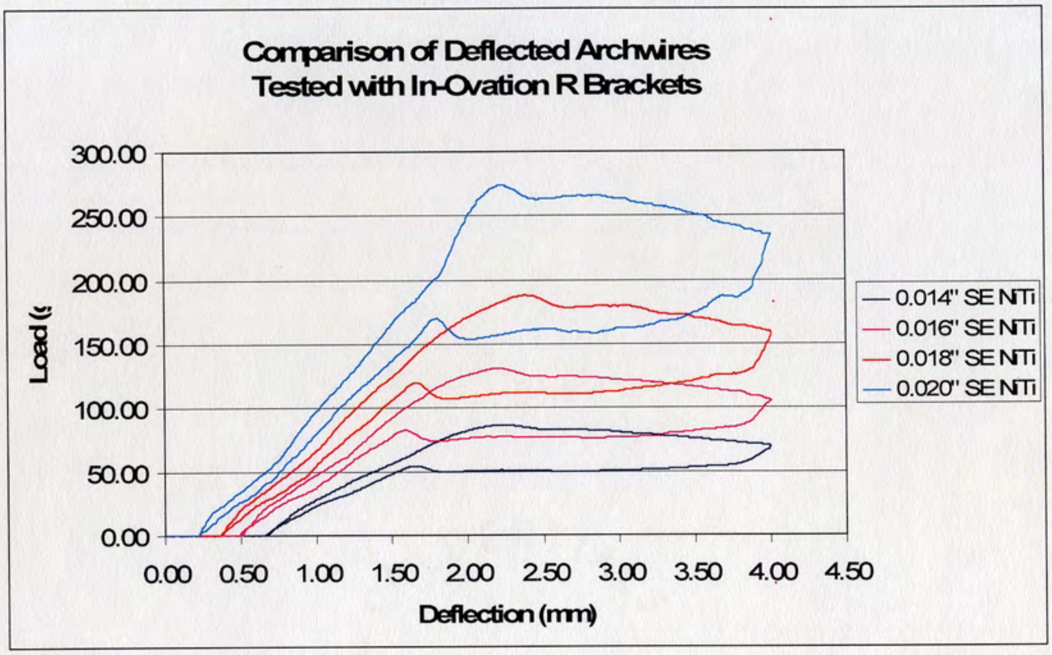




Comparison of Deflected Archwires
Tested with Damon MX Brackets
300.00
250.00
200.00
150.00
100.00
50.00
0.00
Load (
0.00 0.50 1.00 1.50 2.00 2.50 3.00 3.50 4.00 4.50
Deflection (mm)
0.014" SE NiTi
0.016" SE NiTi
0.018" SE NiTi
0.020" SE NiTi



Comparison of Deflected Archwires
Tested with Carriere Brackets
300.00
250.00
200.00
150.00
100.00
50.00
0.00
Load (
0.00 0.50 1.00 1.50 2.00 2.50 3.00 3.50 4.00 4.50
Deflection (mm)
0.014" SE NiTi
0.016" SE NiTi
0.018" SE NiTi
0.020" SE NiTi


Comparison of Deflected Archwires
Tested with In-Ovation R Brackets
300.00
250.00
200.00
150.00
100.00
50.00
0.00
Load (g
0.00 0.50 1.00 1.50 2.00 2.50 3.00 3.50 4.00 4.50
Deflection (mm)
0.014" SE NiTi
0.016" SE NiTi
0.018" SE NiTi
0.020" SE NiTi

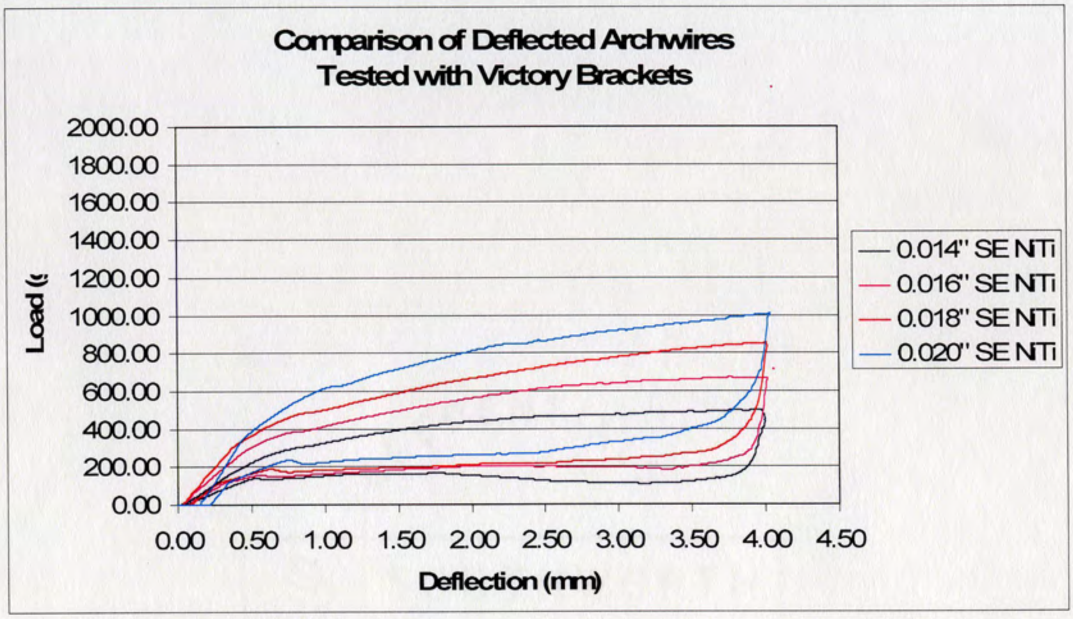
0.014" SE Niti Cantilever Test (4mm Deflection, 5mm Span)								
		Deflection (mm)						**Deflection**
		0.50	**1.00**	**1.50**	**2.00**	**3.00**	**4.00**	**5g Force**
InOvation-R								
		-0.40	22.74	48.47	50.48	50.74	67.22	0.705
		-0.34	23.25	48.61	50.51	49.84	67.69	0.710
		-0.61	22.31	47.43	49.8	50.14	68.49	0.729
	AVG	-0.45	22.77	48.17	50.26	50.24	67.80	0.715
	SD	0.14	0.47	0.64	0.40	0.46	0.64	**8.13**
	% Error	18.48	1.19	0.77	0.46	0.53	0.55	
Carriere								
		0.17	0.135	17.21	48.91	47.76	63.95	1.330
		0.10	0.083	20.22	51.23	48.93	64.81	1.310
		0.27	0.169	18.92	51.11	48.95	65.81	1.320
	AVG	0.18	0.13	18.78	50.42	48.55	64.86	1.320
	SD	0.09	0.04	1.51	1.31	0.68	0.93	**14.79**
	% Error	28.42	19.39	4.64	1.50	0.81	0.83	
SmartClip								
		0.00	-0.16	36.99	46.57	46.92	62.45	1.070
		-0.17	0.065	36.39	45.54	45.97	62.17	1.060
		-0.07	-0.34	36.15	45.72	46.38	62.26	1.080
	AVG	-0.08	-0.15	36.51	45.94	46.42	62.29	1.070
	SD	0.09	0.20	0.43	0.55	0.48	0.14	**12.08**
	% Error	61.66	80.80	0.68	0.69	0.59	0.13	
Damon MX								
		0.51	0.38	18.82	52.65	48.88	63.42	1.340
		0.44	0.64	20.53	54.05	50.48	70.53	1.297
		0.36	0.17	19.42	53.95	50.25	68.89	1.286
	AVG	0.44	0.40	19.59	53.55	49.87	67.61	1.308
	SD	0.08	0.24	0.87	0.78	0.87	3.72	**14.66**
	% Error	9.92	34.27	2.56	0.84	1.00	3.18	
Victory								
		50.45	56.67	57.61	57.09	55.91	64.01	0.047
		48.98	55.42	59.71	55.4	53.71	67.93	0.054
		53.76	58.8	60.64	58.74	57.98	69.54	0.047
	AVG	51.06	56.96	59.32	57.08	55.87	67.16	0.049
	SD	2.45	1.71	1.55	1.67	2.14	2.84	**0.57**
	% Error	2.77	1.73	1.51	1.69	2.21	2.45	

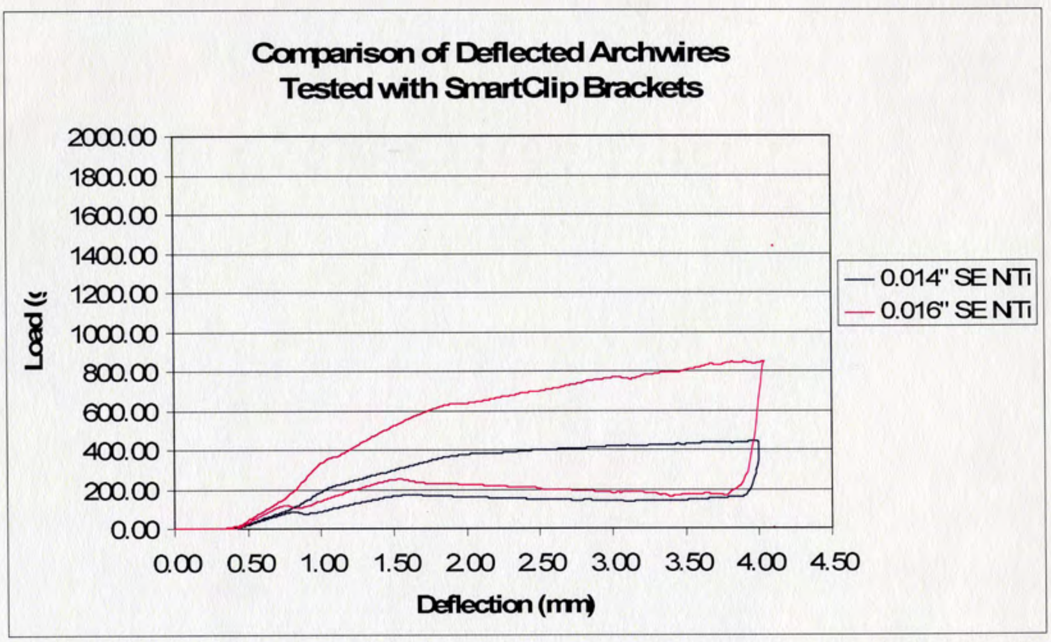
0.016" SE NiTi Cantilever Test (4mm Deflection, 5mm Span)								
		Deflection (mm)						Deflection
		0.50	**1.00**	**1.50**	**2.00**	**3.00**	**4.00**	**5g Force**
InOvation-R								
		-0.56	36.99	74.56	73.47	74.36	104.51	0.510
		-0.51	36.49	77.07	75.04	75.80	106.43	0.540
		-0.42	37.05	76.18	76.30	76.67	104.32	0.534
	AVG	-0.50	36.84	75.94	74.94	75.61	105.09	0.528
	SD	0.07	0.31	1.27	1.42	1.17	1.17	**6.03**
	% Error	7.80	0.48	0.97	1.09	0.89	0.64	
Carriere								
		0.20	0.17	48.14	69.33	71.77	98.74	1.163
		0.30	0.17	49.06	67.79	69.50	97.67	1.155
		0.17	0.10	48.30	70.23	70.92	97.40	1.162
	AVG	0.22	0.15	48.50	69.12	70.73	97.94	1.160
	SD	0.07	0.04	0.49	1.23	1.15	0.71	**13.06**
	% Error	17.60	15.75	0.59	1.03	0.94	0.42	
SmartClip								
		-0.34	15.11	70.35	68.67	68.84	94.22	0.882
		-0.34	16.92	73.74	71.67	72.06	99.71	0.919
		-0.14	15.25	73.12	70.84	71.38	98.41	0.949
	AVG	-0.27	15.76	72.40	70.39	70.76	97.45	0.917
	SD	0.12	1.01	1.81	1.55	1.70	2.87	**10.39**
	% Error	24.39	3.69	1.44	1.27	1.38	1.70	
Damon MX								
		0.36	0.51	47.28	67.98	68.27	98.67	1.180
		0.35	0.68	50.71	69.87	72.61	99.49	1.120
		0.58	0.34	53.66	69.42	71.89	103.12	1.160
	AVG	0.43	0.51	50.55	69.09	70.92	100.43	1.153
	SD	0.13	0.17	3.19	0.99	2.33	2.37	**12.99**
	% Error	17.45	19.25	3.65	0.83	1.89	1.36	
Victory								
		71.50	87.19	83.12	88.16	87.87	102.32	0.042
		77.76	80.57	81.75	81.69	82.39	105.38	0.048
		73.55	82.43	84.86	84.36	86.68	107.92	0.056
	AVG	74.27	83.40	83.24	84.74	85.65	105.21	0.049
	SD	3.19	3.41	1.56	3.25	2.88	2.80	**0.56**
	% Error	2.48	2.36	1.08	2.22	1.94	1.54	

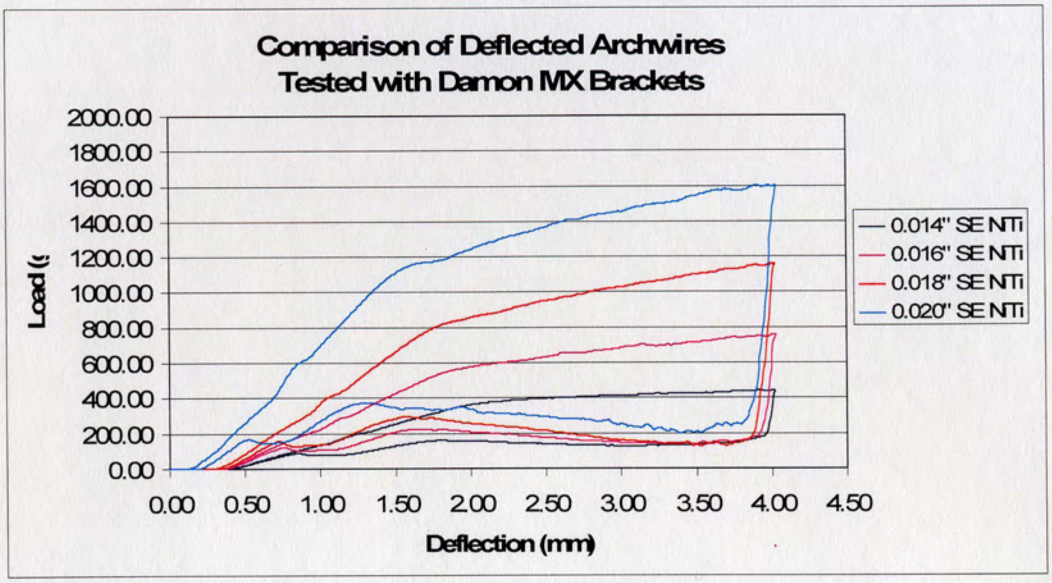
0.018" SE NiTi Cantilever Test (4mm Deflection, 5mm Span)								
		Deflection (mm)						**Deflection**
		0.50	**1.00**	**1.50**	**2.00**	**3.00**	**4.00**	**5g Force**
InOvation-R								
		13.32	55.74	107.12	108.51	114.47	154.95	0.382
		13.24	55.02	106.98	110.56	114.32	155.38	0.398
		12.21	53.74	104.58	108.05	114.55	152.51	0.403
	AVG	12.92	54.83	106.23	109.04	114.45	154.28	0.394
	SD	0.62	1.01	1.43	1.34	0.12	1.55	**4.51**
	% Error	2.76	1.07	0.78	0.71	0.06	0.58	
Carriere								
		0.03	0.44	86.29	83.33	88.13	129.90	1.025
		0.00	1.52	88.87	85.25	88.34	128.78	1.015
		0.07	0.55	89.75	88.04	90.01	130.62	1.018
	AVG	0.03	0.84	88.30	85.54	88.83	129.77	1.019
	SD	0.04	0.59	1.80	2.37	1.03	0.93	**11.52**
	% Error	60.83	40.97	1.18	1.60	0.67	0.41	
SmartClip								
		-0.17	36.59	96.90	94.10	92.91	129.01	0.810
		-0.19	38.96	100.35	96.54	94.90	136.45	0.799
		-0.32	35.33	100.22	95.53	92.36	129.66	0.802
	AVG	-0.23	36.96	99.16	95.39	93.39	131.71	0.804
	SD	0.08	1.84	1.96	1.23	1.34	4.12	**9.13**
	% Error	20.75	2.88	1.14	0.74	0.83	1.81	
Damon MX								
		0.34	0.28	90.81	86.95	88.30	129.86	1.025
		0.44	0.41	92.62	87.79	91.13	131.04	1.005
		0.45	0.59	92.09	90.25	91.21	133.08	1.037
	AVG	0.41	0.43	91.84	88.33	90.21	131.33	1.022
	SD	0.06	0.16	0.93	1.71	1.66	1.63	**11.56**
	% Error	8.57	21.06	0.58	1.12	1.06	0.72	
Victory								
		71.01	107.16	104.86	105.55	109.08	146.55	0.039
		68.04	103.12	101.12	100.26	104.20	139.92	0.046
		67.64	103.47	103.50	104.73	107.75	139.27	0.059
	AVG	68.90	104.58	103.16	103.51	107.01	141.91	0.048
	SD	1.84	2.24	1.89	2.85	2.52	4.03	**0.55**
	% Error	1.54	1.24	1.06	1.59	1.36	1.64	

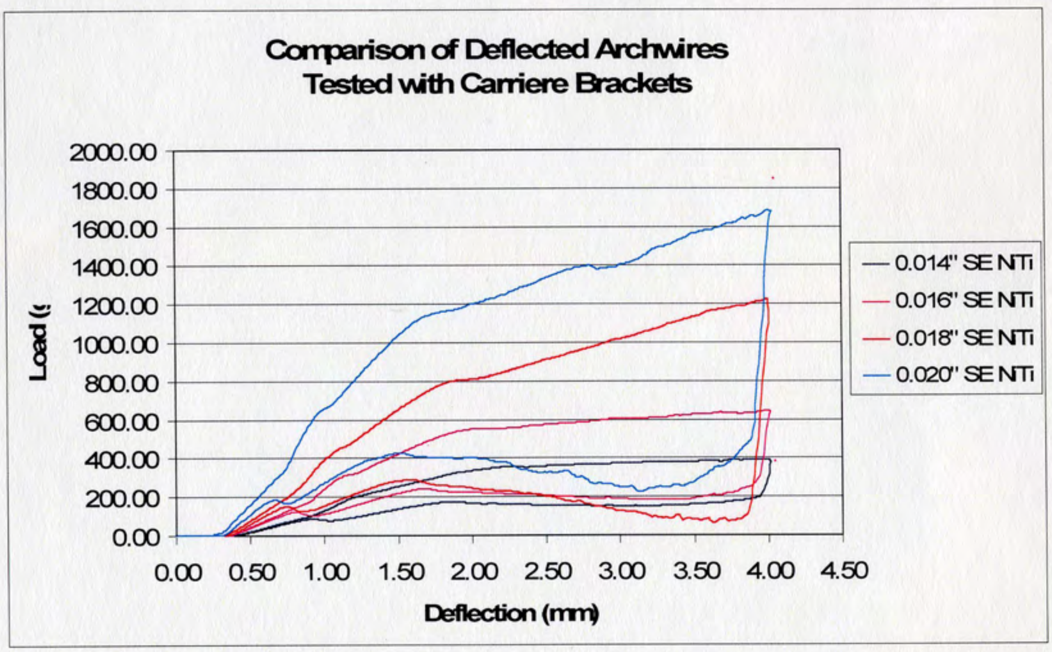
0.020" SE NiTi Cantilever Test (4mm Deflection, 5mm Span)								
		Deflection (mm)						**Deflection**
		0.50	**1.00**	**1.50**	**2.00**	**3.00**	**4.00**	**5g Force**
InOvation-R								
		26.20	83.04	136.98	150.60	159.29	229.60	0.185
		25.88	81.24	135.25	152.96	159.99	232.20	0.239
		26.06	83.60	140.33	156.42	161.09	230.60	0.227
	AVG	26.05	82.63	137.52	153.33	160.12	230.80	0.217
	SD	0.16	1.23	2.58	2.93	0.91	1.31	**2.49**
	% Error	0.36	0.86	1.08	1.10	0.33	0.33	
Carriere								
		0.27	53.24	156.67	146.50	147.92	206.20	0.812
		0.19	49.00	157.80	146.63	148.88	209.10	0.830
		0.10	51.79	153.73	146.62	147.35	210.40	0.830
	AVG	0.19	51.34	156.07	146.58	148.05	208.57	0.824
	SD	0.09	2.16	2.10	0.07	0.77	2.15	**9.36**
	% Error	26.31	2.42	0.78	0.03	0.30	0.60	
SmartClip								
		-0.34	86.12	154.44	154.26	151.77	207.80	0.638
		-0.14	99.60	154.99	161.70	158.89	214.80	0.686
		-0.19	92.87	149.12	157.49	156.70	212.60	0.654
	AVG	-0.22	92.86	152.85	157.82	155.79	211.73	0.659
	SD	0.10	6.74	3.24	3.73	3.65	3.58	**7.51**
	% Error	26.91	4.19	1.22	1.36	1.35	0.98	
Damon MX								
		0.67	46.48	149.01	144.46	145.31	209.60	0.840
		0.34	50.66	158.07	151.18	149.90	211.60	0.825
		0.65	43.46	154.18	150.22	149.01	215.80	0.862
	AVG	0.55	46.87	153.75	148.62	148.07	212.33	0.842
	SD	0.19	3.62	4.55	3.63	2.43	3.16	**9.56**
	% Error	19.31	4.45	1.71	1.41	0.95	0.86	
Victory								
		47.81	66.77	89.07	116.33	152.62	221.10	0.074
		40.55	66.98	88.48	114.15	151.23	218.90	0.133
		42.40	58.96	81.99	101.94	157.06	217.70	0.121
	AVG	43.59	64.24	86.51	110.81	153.64	219.23	0.109
	SD	3.77	4.57	3.93	7.76	3.05	1.72	**1.25**
	% Error	5.00	4.11	2.62	4.04	1.14	0.45	

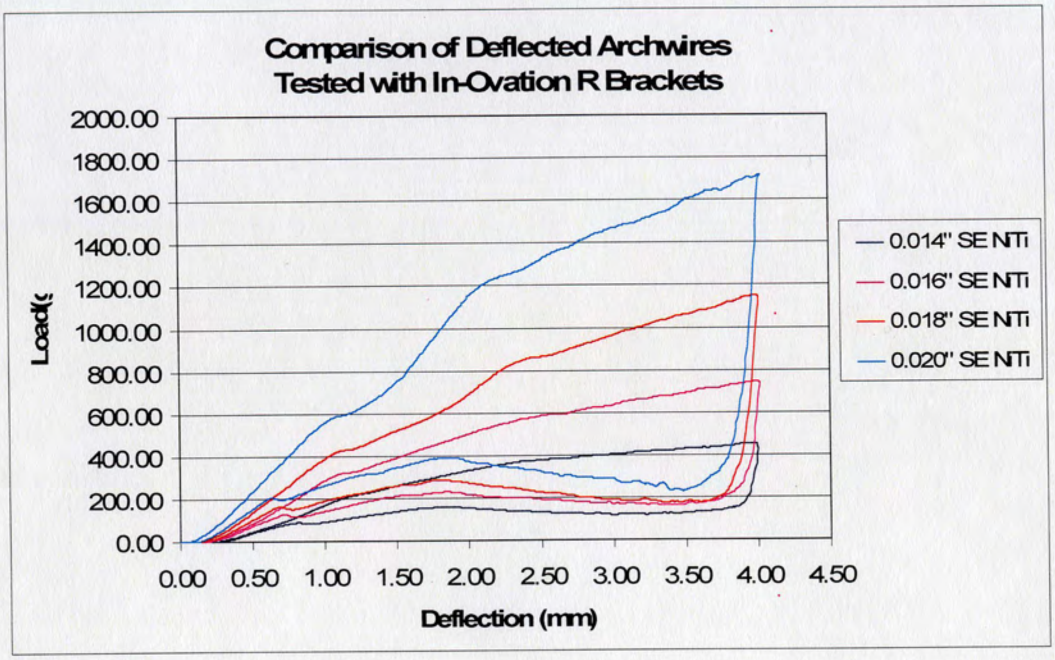
Three Point Bend Graphs






Comparison of Deflected Archwires
Tested with Damon MX Brackets
2000.00
1800.00
1600.00
1400.00
1200.00
1000.00
800.00
600.00
400.00
200.00
0.00
0.00
0.50
1.00
1.50
2.00
2.50
3.00
3.50
4.00
4.50
Deflection (mm)
Load (g
0.014" SE NiTi
0.016" SE NiTi
0.018" SE NiTi
0.020" SE NiTi


Comparison of Deflected Archwires
Tested with Carriere Brackets
2000.00
1800.00
1600.00
1400.00
1200.00
1000.00
800.00
600.00
400.00
200.00
0.00
0.00
0.50
1.00
1.50
2.00
2.50
3.00
3.50
4.00
4.50
Deflection (mm)
Load (g
0.014" SE NiTi
0.016" SE NiTi
0.018" SE NiTi
0.020" SE NiTi


Comparison of Deflected Archwires
Tested with In-Ovation R Brackets
Load(g
Deflection (mm)
0.014" SE NiTi
0.016" SE NiTi
0.018" SE NiTi
0.020" SE NiTi
2000.00
1800.00
1600.00
1400.00
1200.00
1000.00
800.00
600.00
400.00
200.00
0.00
0.00
0.50
1.00
1.50
2.00
2.50
3.00
3.50
4.00
4.50

0.014" SE NiTi 3pt Bend Test (4mm Deflection, 10mm Span)								
		Deflection (mm)						**Deflection**
		0.50	**1.00**	**1.50**	**2.00**	**3.00**		**5g Force**
InOvation-R								
		39.72	93.19	144.07	156.33	120.31		0.294
		39.25	86.64	134.77	151.81	119.41		0.280
		37.83	87.92	135.29	148.3	116.84		0.299
	AVG	38.93	89.25	138.04	152.15	118.85		0.291
	SD	0.984	3.472	5.226	4.026	1.801		**3.33**
Carriere								
		16.64	77.98	140.71	169.78	158.38		0.398
		14.64	79.98	135.37	167.06	152.27		0.423
		15.86	78.06	131.22	162.17	146.15		0.431
	AVG	15.71	78.67	135.77	166.34	152.27		0.417
	SD	1.008	1.132	4.757	3.856	6.115		**4.77**
SmartClip								
		20.39	90.80	168.31	162.82	141.84		0.404
		19.13	86.40	161.79	155.11	140.26		0.417
		18.16	83.77	162.40	164.74	135.11		0.414
	AVG	19.22	86.99	164.17	160.89	139.07		0.412
	SD	1.120	3.552	3.601	5.097	3.519		**4.71**
Damon MX		17.73	78.43	130.21	159.35	124.71		0.411
		15.95	78.02	127.22	154.97	131.23		0.415
		16.78	76.50	125.49	156.01	127.44		0.422
	AVG	16.82	77.65	127.64	156.78	127.79		0.416
	SD	0.891	1.017	2.388	2.288	3.274		**4.76**
Victory		143	151.26	163.11	152.76	99.93		0.014
		135.7	154.46	168.57	157.85	119.84		0.070
		130.16	149.74	158.23	152.12	116.01		0.073
	AVG	136.29	151.82	163.30	154.24	111.93		0.052
	SD	6.440	2.409	5.173	3.140	10.564		**0.60**

0.016" SE NiTi 3pt Bend Test (4mm Deflection, 10mm Span)								
		Deflection (mm)						**Deflection**
		0.50	**1.00**	**1.50**	**2.00**	**3.00**		**5g Force**
InOvation-R								
		68.31	131.27	198.87	200.70	168.94		0.206
		68.62	138.06	209.30	215.60	170.46		0.201
		67.38	137.86	201.30	210.80	162.00		0.214
	AVG	68.10	135.73	203.16	209.03	167.13		0.207
	SD	0.645	3.864	5.457	7.605	4.510		**2.37**
Carriere								
		36.12	110.61	214.00	219.90	174.07		0.372
		38.74	112.33	216.40	217.60	187.02		0.379
		39.70	111.03	217.80	223.70	188.30		0.375
	AVG	38.19	111.32	216.07	220.40	183.13		0.375
	SD	1.853	0.897	1.922	3.081	7.872		**4.29**
SmartClip		48.42	150.36	247.90	225.40	201.60		0.36
		45.46	142.44	248.50	228.60	188.70		0.361
		42.36	142.82	247.60	219.30	189.07		0.358
	AVG	45.41	145.21	248.00	224.43	193.12		0.360
	SD	3.030	4.467	0.458	4.725	7.343		**4.11**
Damon MX								
		33.57	107.52	210.60	217.70	175.57		0.368
		29.72	105.94	208.00	208.50	161.97		0.384
		34.67	105.61	205.00	206.80	145.08		0.361
	AVG	32.65	106.36	207.87	211.00	160.87		0.371
	SD	2.599	1.021	2.802	5.864	15.275		**4.24**
Victory								
		148.55	162.01	184.68	199.43	192.48		0.024
		146.98	162.86	184.37	200.80	194.16		0.038
		146.41	158.30	171.77	187.90	178.22		0.040
	AVG	147.31	161.06	180.27	196.04	188.29		0.034
	SD	1.108	2.425	7.366	7.086	8.758		**0.39**

0.018" SE NiTi 3pt Bend Test (4mm Deflection, 10mm Span)								
		Deflection (mm)						**Deflection**
		0.50	**1.00**	**1.50**	**2.00**	**3.00**		**5g Force**
InOvation-R								
		97.54	188.38	259.80	280.30	200.40		0.165
		96.84	183.09	252.90	256.50	178.59		0.164
		100.18	194.19	258.20	269.60	183.92		0.154
	AVG	98.19	188.55	256.97	268.80	187.64		0.161
	SD	1.761	5.552	3.612	11.920	11.370		**1.84**
Carriere								
		57.89	155.35	300.40	252.40	141.71		0.334
		54.17	144.54	273.10	237.30	108.36		0.346
		57.76	156.11	271.80	244.50	130.76		0.340
	AVG	56.61	152.00	281.77	244.73	126.94		0.340
	SD	2.111	6.472	16.150	7.553	16.999		**3.89**
SmartClip								
	Archwire Released due to High Force @ ~3.5mm							
Damon MX								
		54.62	137.62	290.00	269.10	164.06		0.317
		57.02	133.77	287.00	254.20	156.59		0.318
		54.29	133.50	272.70	242.30	156.56		0.327
	AVG	55.31	134.96	283.23	255.20	159.07		0.321
	SD	1.490	2.305	9.245	13.428	4.321		**3.67**
Victory								
		160.10	178.75	194.30	205.30	224.80		0.037
		170.24	186.60	202.70	216.10	255.00		0.033
		159.22	179.92	191.29	207.50	222.50		0.055
	AVG	163.19	181.76	196.10	209.63	234.10		0.042
	SD	6.124	4.235	5.913	5.707	18.136		**0.48**

0.020" SE NiTi 3pt Bend Test (4mm Deflection, 10mm Span)								
		Deflection (mm)						**Deflection**
		0.50	**1.00**	**1.50**	**2.00**	**3.00**		**5g Force**
InOvation-R								
		159.86	277.30	377.10	393.20	322.10		0.081
		158.06	272.20	361.40	381.20	300.60		0.075
		150.67	261.40	344.40	370.70	289.40		0.087
	AVG	156.20	270.30	360.97	381.70	304.03		0.081
	SD	4.870	8.118	16.354	11.258	16.618		**0.93**
Carriere								
		101.03	272.30	447.30	385.80	299.10		0.269
		107.19	276.30	444.50	365.60	285.60		0.260
		105.13	270.50	425.30	401.30	249.30		0.259
	AVG	104.45	273.03	439.03	384.23	278.00		0.263
	SD	3.136	2.969	11.976	17.901	25.755		**3.01**
SmartClip								
	Archwire Released due to High Force @ ~3.5mm							
Damon MX								
		164.43	291.30	369.30	359.80	259.70		0.138
		157.31	273.40	351.20	318.10	227.70		0.146
		159.82	266.40	346.50	337.70	249.40		0.142
	AVG	160.52	277.03	355.67	338.53	245.60		0.142
	SD	3.611	12.841	12.038	20.862	16.335		**1.63**
Victory								
		136.65	211.70	226.60	256.30	317.30		0.003
		186.72	231.70	255.60	275.30	334.50		0.096
		167.16	223.10	246.10	263.10	330.90		0.152
	AVG	163.51	222.17	242.77	264.90	327.57		0.083
	SD	25.234	10.033	14.785	9.627	9.072		**0.96**