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Review

Understanding the Force Deflection Behavior of NiTi Archwire at Distinct Bending Configuration: A Narrative Review in Vitro Studies

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Abstract. This study aims to assess the optimal unloading force range for human comfort by considering NiTi archwires in different bending settings, based on previous research findings. All the relative data has been collected from different databases such as PubMed, Google Scholar, Scopus, Web of Science, and USM library. The publications from 2007 till February 2023 have been incorporated. Several parameters related to orthodontics, especially the usage of three brackets and three-point bending with respect to optimal tooth force were taken into consideration. These parameters, however, included various aspects like the shape memory effect, bending temperature, friction, and gingival/labial direction. ISO standards pertaining to the bending tests were also contemplated in this review. The study examined 74 articles related to orthodontic tooth movement, three brackets, and three-point bending. In fact, this review was done to analyze the force deflection behavior and related parameters to orthodontics. For this, among 74 selected research items, 15 studies gave information about the optimal tooth force, 8 focused on the optimal ranges, while 7 reports indicated the higher rates of tooth force. All these studies illustrated the considerable variation in methodology and clinical diversity in terms of applied forces. This article summarizes previous investigations on orthodontic tooth force, highlighting the ideal range of 0.2 to 1.5 N. It concludes that maximum force decreases with greater inter-bracket distance but increases with wire deflection and testing temperature. Proper force management is emphasized as crucial for preventing unwanted tooth movement and its biological consequences.

Keywords: NiTi archwire, force deflection, bending configuration, orthodontics.

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1. Introduction

Nickel-Titanium alloys, because of their unique features, excellent mechanical capabilities, biocompatibility and corrosion resistance, are commonly used as orthodontic wires. When multiple factors such as bending temperature, bracket friction, inter bracket distances, wire size and wire geometry are taken into account, these wires have become the favored choice for both dentists and patients [1]. These alloys have three subdivisions namely conventional alloy, superelastic alloy and thermoelastic alloy [2]. The superelasticity of NiTi alloys and the elasticity of human bones, in addition to the viability of elasticity and strain shielding, must all be assessed in order to determine the appropriateness of using these wires in orthodontics [3].

The deflection of the orthodontic wire is proportional to the magnitude of the applied force, and a recent study concluded that the plateau slope increased with increasing temperature up to a particular limit of 26 °C to 46 °C [4]. Nowadays, "light continuous forces" are considered physiologically acceptable and effective; however, the concept is applied somewhat subjectively in this context. Indeed, the scientific community has established no consensus on what constitutes a "light" force and what does not. Thus, there is no definite scientific evidence quantifying the best force for orthodontic movement, and clinicians must determine the force that is most appropriate for each clinical circumstance.

Frictional force is a critical counterbalance parameter for tooth movement. Recently, a study was conducted to determine the usefulness of artificial saliva. It had been demonstrated that TiNiCu wires exhibit a significant increase in static and kinetic forces during dry and wet test conditions (artificial saliva) while friction decreases in general when wet conditions [5]. The diameter of the archwire and the type of ligation used influence the misalignment correction. Archwires with a smaller diameter are recommended for levelling and alignment to minimize excessive stresses that could create undesired side effects in the therapeutic setting [6]. It is also evident that the type of material influences the frictional force regardless of moisture or temperature. Therefore, material selection is also critical for optimizing the orthodontic procedure [5].

The effect of temperature, duration of NiTi ageing solution treatment on the chemical composition of the archwire and the austenite to martensite transformation phase improves the efficiency of NiTi wire. After twenty hours of annealing at 450-600 °C, the grain structure is completely lost; however, after two hours at 900 °C the grain structure was then visible. Recrystallization is not responsible for grain structure degradation because grains had non-random distributions [7].

This gradient property of stress-induced phase change accounts for the progressive rise in dragging force which is proportional to the size of the patient's teeth. 400 to 460 °C is the ideal temperature to induce gradient stress plateau stress-induced phase transformation of Ti-50.8 at 1 % Ni for austenite to martensite transformation. Additionally, it was noted that treatment with gradient temperature produced a positive and virtually linear stress plateau consistent with 130 MPa stress for both phase transitions [8]. However, by carefully selecting the ageing time and temperature on pseudo elasticity under high stress, the mechanical properties of wire can be improved. Thus, following unloading, deformation is almost completely recovered with less than 0.5 % irrecoverable strain [9].

The appropriate amount of force to apply when moving orthodontic teeth movement (OTM) is quite ambiguous Therefore, orthodontic operations may take longer than expected and have unpleasant side effects [10]. Orthodontic mechanotherapy alters paradental tissues to shift teeth. It could only require 0.196 to 1.47 N per tooth [11]. The ideal orthodontic forces for canine distal-direction translational and tipping motions have been calculated to be between 1.27 and 1.34 N in recent studies [12]. The optimal tooth force, which is estimated to be between 0.2 and 1.5 N on average, is calculated by taking into consideration the results of prior studies.

According to our most recent research, the short-term aged NiTi archwire could potentially be employed to improve the force delivery tendency to the misaligned tooth by reducing the force's magnitude and maintaining it throughout the orthodontic treatment period [13]. The prime motivation for this study was to assess NiTi wires that have been bent in a wide range of configurations, such as shape memory effect, geometry of wire, type of bracket, inter bracket distance, binding friction, and bending temperature. The unloading force (tooth movement) must be far lower than the indicated ideal force ranges for human comfort based on the findings of past studies.

2. Three-Point Bending

The mechanical properties of archwires are normally evaluated using a 3-point bending test to establish whether orthodontic force delivery is optimal, predictable, and successful during orthodontic treatment [14]. Usually three-point bending test is conducted in a way that, a load cell is brought into contact with a section of wire and deflected in line with wire according to the orthodontic wire standard ISO 15841 and at predetermined intervals, loading (ligating) and unloading (tooth-moving) forces and the wire's permanent deformation values would be measured [15]. In earlier studies, it was performed in buccolingual plane, analogous to first-order wire deflection in a universal bending machine. Vertical force can be applied to the midpoint of the wire between the central incisor and canine teeth by a rod attached to the machine's moving head at a crosshead speed of 1 mm/min for a deflection of 2 mm. The unloading phase proceeded at the same rate and the force was measured during loading and unloading. The same test protocol would be used to determine the severity of malocclusion (4 mm wire deflection) [16].

According to results from prior research during the 3-point bending test, the lowest force generated upon activation of 0.016 inch round wire at 1 mm was 95 ± 10 g, while the highest force generated at 3 mm was 165 ± 10 g. The minimum activation force in 0.016 × 0.022 inch rectangular wire was 210 ± 10, while the highest force was 340 ± 10 at 3 mm deflection. For deactivation, the minimum for 0.016 wire at 1 mm deflection was 150 g, but the highest force was 295 g at 3 mm. For the 0.016 × 0.022 inch wire, the lowest force at 1 mm deflection was 150 ± 20 g and highest force was 295 g [17]. Interestingly, a recent investigation concluded that both as received and retrieved bio-force NiTi archwires had several force zones. These archwires' austinite finish temperature (A_f) temperatures may be greater than the normal intraoral temperature. In comparison to retrieved specimens, higher forces were reported during the three-point bending test in as received wires. The forces recorded from these archwires may exceed biologically acceptable limits even at 2 mm deflection [18].

2.1. ISO Standards Related to Orthodontic Application

ISO standards provide standard specifications and testing procedures for orthodontic wires in fixed and removable appliances. These standards specify physical and mechanical attributes, test procedures, packaging, and labeling information. While they do not have specific requirements for biological risks, it is recommended to reference ISO 7405 and ISO 10993-1 when assessing potential biological concerns. Bending tests are conducted to determine elastic modulus, validation strength, elongation after fracture, and tensile testing calibration for crosshead rates ranging from 0.5 mm/min to 2.0 mm/min [19]. Figure 1 depicts the three-point bending test setup and loading and unloading forces applied to a NiTi wire during testing. According to ISO 15841, for Nitinol heat-activated archwire, at the unloading of a three-point bending test with a span of 10 mm at a temperature of 36 ± 1 °C and austenite finish temperature range of 20 to 40 °C, the maximum permanent deflection should be 2 % [20].

2.2. Shape Memory Effect of NiTi Wires

Shape Memory Alloys (SMAs) have attracted a lot of attention and interest in recent years in a wide range of commercial applications due to their unique and superior features and this commercial development has been backed up by basic and applied research studies [21]. Most importantly, NiTi SMAs offer versatile applications in orthodontics, orthopedics, and cardiovascular surgery due to their unique shape memory effect and pseudo elasticity. This eliminates the need for external intervention, such as open surgery, in device adjustments and repositioning [22], [23]. However, the characteristics of these alloys vary according to composition, transformation temperature range, and loading & unloading forces [24].

The transformation temperatures can be tailored and superelastic recovery of up to 5.5 % can be achieved at body and room temperatures by following thermal treatments of 350 °C for 1 hour and 600 °C for 1.5 hours. These alloys are extremely promising for potential biomedical applications based on full-field strain measurements. The underlying martensitic phase transformation morphologies evolve uniformly for the standardized Selective Laser Melting (SLM) alloy microstructure, whereas localized strain concentrations emerge for laser-based direct energy deposition (LDED). Deformation study indicated that the SME recovered roughly 2 % macro-scale and 4 % micro-scale stresses [25].

The NiTi alloy has 50.8 Ni and 49.2 % Ti mixed for 12 hours using sintering process, has lower density, higher modulus, and similar strength as dense NiTi shape memory alloy. It displays shape memory effect and stable linear superelasticity, but properties vary based on composition, transformation temperature, and loading/unloading forces [5], [26].It has been observed in prior study that the factor of ageing could be a disadvantage for Cu-based SMAs, as it limits their application to low-temperature situations due to their predisposition for phase stabilization, transformation temperature hysteresis and shape memory effect degradation at relatively high temperatures. The irregularity and degradation of SME in copper based SMAs have been linked to ageing processes such as martensite depletion and pinning of the austenite/martensite interface [27].

2.3. Force Deflection Behavior

The force behaviour of NiTi archwires changes depending on the degree of bending, the distances between bracket [28]. The conventional three-point test reveals the horizontal unloading plateau of superelastic NiTi alignment wires. When superelastic NiTi wires were tested with brackets at the wire-testing machine which produce forces with a significant degree of qualitative and quantitative variability, which was dependent on the extreme wire deflection and on same unloading data point, NiTi wires deflected to substantially different maximum deflections (2 mm and 4 mm). This feature enables therapists to modulate the force released during alignment of the NiTi wire [1].

The magnitude of force delivered by NiTi alloy archwires and brackets when ligated conventionally with an elastic module was greater than the magnitude of force delivered when self-ligated with a slide. Despite the use of diverse ligation procedures, NiTi alloy archwires demonstrated considerable increases in force magnitude on two neighboring teeth with the smallest diameter increase [29]. Interestingly, heat-activated versions usually generate lighter forces over larger deflection plateaus,

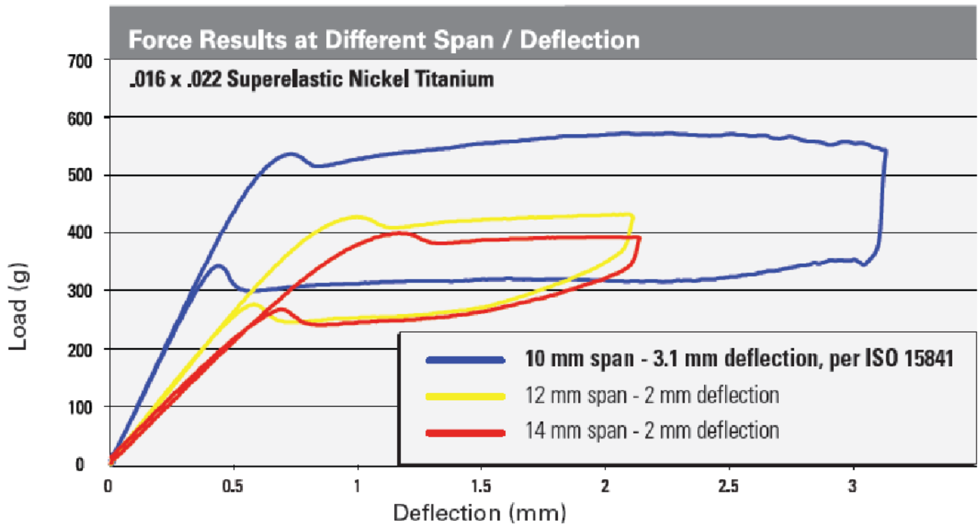

Fig. 1. Three-Point Bend Representative loading and unloading forces on a Nickel Titanium wire during testing [15].

despite the fact that there was significant variance in the plateau behaviour. The differences between conventional and heat-activated wires were noted in this case, on average, the increase in plateau force was about 50 % when the diameter was increased by 0.002 inch (from 0.012 to 0.014 and from 0.014 to 0.016 inch), while about 150 % enhanced in plateau force when it was increased by 0.004 inch [30].

2.4. Thermal-Stress Analysis for 3-Point Bending

The ageing treatment influences both the transformation and deformation behaviour of Ti-50.9 at. % Ni. Increasing the ageing temperature within 673-873 K, the reverse transformation temperature often falls and complicated forward transformation behaviour results [31]. The optimum super-elasticity in three-point bending at body temperature is obtained after 30 minutes of annealing at 300 °C. This results in a reduced, consistent unloading force and a minimal stable set is suitable for orthodontic uses [32].

Interestingly, in our recent study it was observed that the bending behavior of superelastic NiTi archwires was altered by subjecting them to different temperatures in an ageing treatment for 15 minutes. The commercial NiTi archwires showed better bending forces in three-point and three-bracket configurations. Aged at 490 °C or 520 °C, the archwires exhibited lower magnitude and more consistent force during unloading. The 15-minute ageing treatment produced a suitable size of Ni_4Ti_3 precipitate, making the wire more flexible and reducing unloading force in the three-bracket configuration [13].

3. Three-Bracket Bending

A three-bracket bending test can be performed utilizing an Instron universal testing machine (UTM) at a temperature of 37 °C, simulating the oral environment and frequent clinical conditions and procedures. In addition to stainless steel and elastomeric ligatures, active and passive self-ligation could be employed; however, because ligation is an important feature of any bracket design, the independent variables were wire diameter and bracket type. During leveling and alignment, it is advisable to utilize equally efficient archwires with a smaller diameter to avoid applying excessive stresses that may create undesirable side effects [6]. To construct complex three-dimensional (3D) tooth motions, fixed multiple-bracket appliances are frequently used. The tooth crowns are covered with several brackets, and an archwire is implanted into the bracket slots and secured with steel or elastic ligatures. The appliance is turned on by introducing archwires in a specific order, progressively increasing the effective wire stiffness (EWS) and tapering the space between the wire and bracket slots [33].

Self-ligating brackets, according to orthodontic distributors, provide numerous benefits, including decreased friction, faster archwire changes, complete archwire engagement in the slot, increased patient comfort and hygiene, fewer emergencies, decreased root resorption, decreased need for extractions due to arch expansion, faster treatments, fewer appointments, better results, and increased efficiency [32].

The mechanical properties of an orthodontic wire can be classified into two categories: intrinsic or true properties, which refer to the wire's true nature, and effective properties, which refer to the wire's true nature when it comes into contact with the bracket or ligation system, the intraoral temperature and its transition etc. [34]. The following bracket specification, i.e., bracket width and free wire length, are expressed in Eq. (1) and (2), respectively.

$$k^{C_1} = \frac{\Delta F}{\Delta \delta^{C_1}} = \frac{12EI}{l^3}\left(\frac{5bl + 6l^2}{6b^2 + 10bl + 3l^2}\right) \tag{1}$$

$$k^{B} = \frac{\Delta F}{\Delta \delta^{B}} = \frac{12EI}{l^3}\left(\frac{6l}{2l + 3b}\right) \tag{2}$$

where the variables b and l specify the width and length of the brackets and F and E denote force and young's

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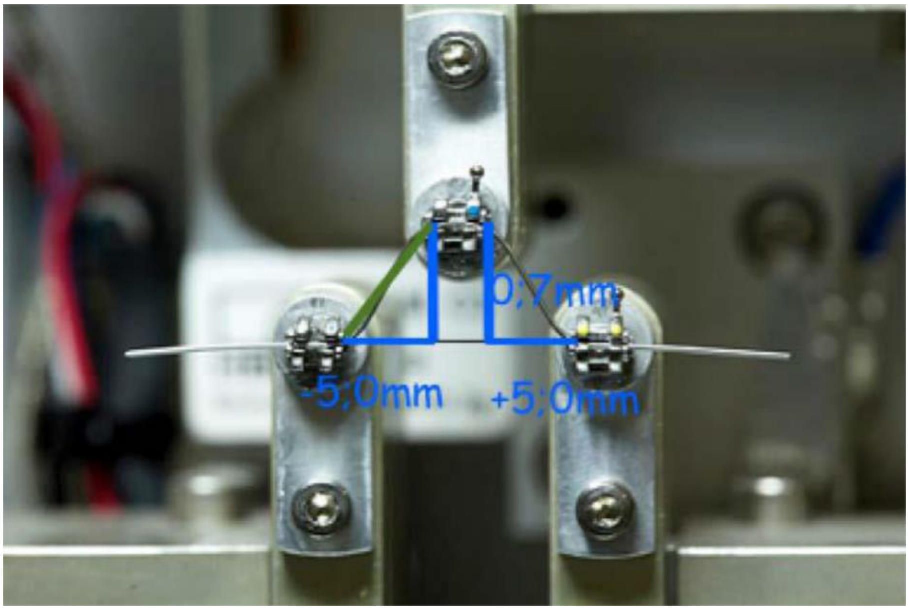

Fig. 2. A right triangle with catheti of 5 mm and 7 mm was constructed [36].

modulus, respectively. The ratio between bracket width and free wire length (k^{C_1}/k^{B}) determines the effective wire stiffness (EWS). After modification for edge radius, the ratio reveals a potential insincere of EWS by 301 %. The absence of slot height in this equation implies that this variable is irrelevant in this equation [33]. Achieving an acceptable incisor inclination or torque is critical for the ultimate cosmetic result. Torque expression is dependent on a variety of elements, including the slot size and archwire diameter of the bracket. The complete torque expression is possible by utilizing an archwire of the suitable size to fill the bracket slot; however, in order to enter a full size rectangular archwire, a certain degree of 'play' is required. This indicates that the bracket slot's vertical dimension or height must be greater than the archwire's height [35].

It is widely established that friction between the wire and the bracket can impair the efficiency of tooth movement during the leveling step. Numerous self-ligating brackets have been created and sold under the term's low friction or friction-less. This is a commonly known advantage of self-ligating brackets [34]. Most companies now offer self-ligating brackets with low-stretch wires to facilitate tooth movement with less friction and force. Figure 2 depicts a right triangle with catheti bracket specifications such as bracket height and bracket distances reference to the central incisor and premolar, etc. The data on the premolar and lateral incisor brackets reveal a decreasing horizontal force (101-13 cN) and increasing vertical force (10-51 cN), as well as a constant moment (378 nNmm - 310 cNmm). Additionally, it indicates that, while the horizontal force and moment acting on the canine bracket are minimal, the vertical force acting on it (19-108 cN) is twice as strong as the vertical force acting on the premolar and lateral incisor brackets [36].

3.1. Wire Size and Geometry of Archwire

It is widely accepted that an archwire is chosen for a particular clinical condition based on the alloy's mechanical qualities, that are optimally balanced in terms of stability, stiffness, resilience and formability [37]. These archwires are available in two types: fiber-reinforced composite and coated metal. In recent advancements, a sensor was attached to each bracket and wired passively. After carefully adjusting the location of each sensor, the original force and moment were set to zero [38]. As seen in Fig. 3, the testing jig was set up with typical twin brackets and a sensor glued to a bracket, and wire was applied passively.

The hysteresis loop of the material and friction between both the archwire and the bracket causes the vertical difference between two curves. The loading curve depicts the force required to engage the wire in the bracket, whereas the unloading curve depicts the amount of applied force to the teeth throughout the levelling plus aligning process [38]. The cross-section of NiTi heat activated archwires had substantial influence on force levels. When 0.016 inch archwires were used instead of 0.014 inch archwires, the biggest increase in force level occurred with self-ligating brackets, whereas the lowest increase occurred with traditional brackets ligated using elastomeric rings [39]. On the other hand, the bracket design has a significant impact on the amount of force exerted by the super-elastic NiTi alignment wires. This type of experiment has a qualitative and quantitative effect on the wire's force release. Only the usual three-point test exposes the super-elastic NiTi alignment wires' horizontal unloading plateau. When superelastic NiTi wires were tested with brackets at the wire-testing machine interface, they exhibited a high degree of qualitative and quantitative variability in their force release, which was dependent on the maximum wire deflection [40]. When longer archwires and more deflection are used, the deactivation force increases. On

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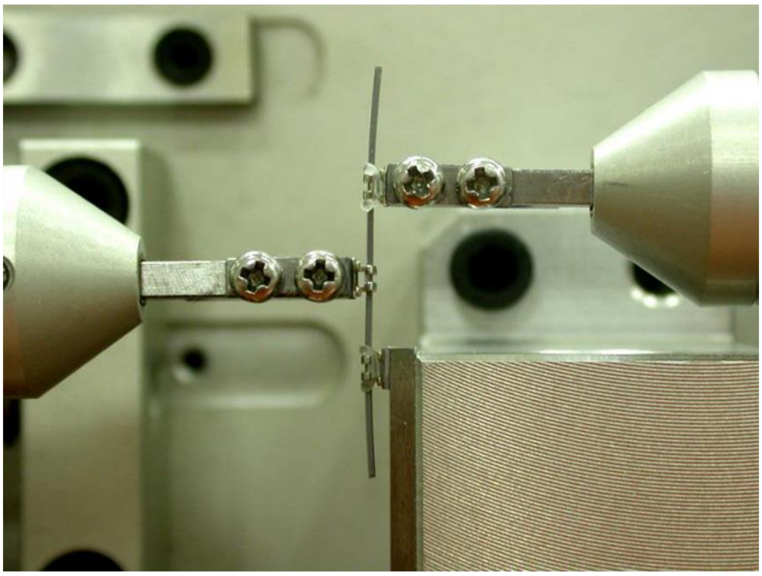

Fig. 3. Sensor bonded with a bracket, and wire was applied passively [42].

the other hand, the deactivation energy is less affected by the diameter of the archwire and the degree of deflection. Also, mechanical efficiency diminishes when greater deflection and larger archwires in size. Moreover, passive, and active self-ligating devices had shown minimal difference in terms of comparative performance based on human comfort parameters [41].

Recent research has shown that the 0.016 inch diameter archwire, with a lighter force range of 0.57 N to 1.71 N and a lower force slope of 0.13 N/mm to 0.72 N/mm, is better suitable for usage during the first phases of orthodontic therapy. On the other hand, the usage of 0.016 × 0.022 inch rectangular wire may put the patient through pain since the force required to move the teeth may be greater than 3.61 [43]. Interestingly a wire's activation and deactivation tendencies could not be the same. The force deflection graphs produced during the activation (loading) and deactivation (unloading) cycles are therefore different. For clinicians to choose the best wire, understanding deactivation behaviour is important. The activation and deactivation forces likewise rise with increasing wire cross-section, from 0.014 to 0.016, regardless of the wire material, as seen in all tables and graphs. According to this, the load-deflection forces are precisely proportional to the wire's cross section. The maximum load-deflection forces were seen for 0.016 nickel-titanium wires during testing, whereas the lowest forces were observed for 0.014 coaxial wires [44].

3.2. Number and Types of Brackets

Orthodontists are concerned with repositioning teeth utilizing therapeutic pressures and moments applied to distinct teeth through detachable or permanent appliances. Multiple-bracket fixed appliances are frequently used to achieve complex three-dimensional (3D) tooth motions. Numerous brackets are bonded to the dental crowns, and an archwire is inserted into bracket slots and secured with elastic or steel ligatures. The appliance is activated by adding archwires successively, progressively increasing the effective wire stiffness (EWS) and decreasing the distance between the wire and bracket slots [33].

Self-ligating brackets are increasingly being employed in place of traditional brackets for a variety of reasons. One of them is, it applies decreased friction when used in conjunction with thinner archwires during the first leveling and alignment step. However, with the advent of NiTi alloys into orthodontics, there has been an increasing trend toward early treatment with bigger continuous archwires [45].

According to research that evaluated the influence of ligation on the load-deflection properties of NiTi wires, the elastomeric ligature (EL) acts as a constraint on superelastic wires. Thus, the findings of this investigation indicated that the predictability of the released force is larger with the metal ligature (ML) and self-ligating system than with elastomeric ligature. The benefit of the self-ligating brackets (SLB) is that they release lighter forces. In substantial deflections, thermally activated NiTi wire showed stronger forces than traditional NiTi wire. Low friction brackets (self-ligating and traditional brackets tied with metal ligature) demonstrated higher force uniformity than traditional brackets with elastomeric ligature. ML enables the application of bigger magnitudes of forces than SLB. In considerable deflection, the active self-ligating system demonstrated lower forces than the passive system [46].

The most advantageous combinations for generating the highest torque moment between the wire–bracket–ligation system for archwire sizes and materials in comparison to the two self-ligating brackets used in the study of [47]. Besides, it is evident that the lowest forces were measured when the brackets were combined with either the coaxial or thermal alloy archwires [48]. According to the prospective investigation, the passive self-ligation approach generated a more accurate result based on the in-vitro data force mechanism for this malocclusion, resulting in less undesired forces and moments as compared to standard elastic ligation [49].

Clinically, the three-bracket model reflects a simplification of the geometrical connections seen in orthodontic fixed-appliance therapy. In practical practice, inter-bracket spacing is not consistent, brackets are improperly angulated, and the wire is ligated to the curved dental arch [33].

3.3. Binding Friction at Wire – Bracket Interfaces

Friction is a fundamental physical phenomenon that must be managed in order to respond predictably each time using prescribed mechanics for the amount of force imparted to the teeth to more closely match the amount of force applied to them and the wire–bracket coupling [50]. The amount of archwire deflection-induced friction, also known as binding, is determined by the amount of force applied by the archwire to the anchorage bracket walls and is controlled by the inter-bracket distance, diameter and type of orthodontic wire alloy [51]. Orthodontic tooth movement is characterized by the degree of staticity and this is particularly true during the first hyalinization phase. Thus, static testing of archwires offers accurate reference data for the peak stresses exerted immediately after clinical insertion of a leveling wire. Due to experimental friction, which may be much different from the friction that encountered during clinical treatment, considerable underestimating of real pressures produced on individual teeth may occur during dynamic wire testing [52].

Self-ligating brackets exhibit significant friction resistance when stainless steel wire is deflected in the buccolingual plane. Here, the coefficient of friction was related to the stiffness of the wire-securing devices and the amount of wire bending since there is a relation between passive clip design in self-ligating brackets and the scratching of the wire surface [53]. Classical friction (FR) occurs because of the ligation force pressing the archwire into the bracket slot's base and wall. The rounded arch slot walls of the synergy brackets utilized in their research have lower BI (Basilar Impression) and FR, which are critical for managing slippage between the archwire and bracket during the leveling process. Additionally, when the elastomeric ligation was secured to the inner tie wings of the synergy brackets, it allowed unrestricted movement of the archwire inside the bracket slot (passive ligation design) and as a result, frictional force was decreased [54].

In prior investigation, it was determined that the wire bent in the bracket model is far more resistant to friction than the wire bent in the point form. When the wire was deflected to 3.0 mm in the bracket model, the largest frictional force was produced, with a magnitude of 2.01 N during loading and 1.61 N during unloading. When the point support was taken into account, the friction values associated with unloading decreased dramatically to 0.25 N [55].

3.4. Effect of Inter-Bracket Distance

Archwire and bracket slot combinations result in a range of applied torque values. As a result, the orthodontist must choose an optimal combination of labial archwire diameters and slot size. The MBT versatile appliance system provides more palatal root torque in the upper incisor region and more labial root torque in the lower incisor region [56], as shown in Fig. 4. The clinical significance depends on model constraints, especially the inter-bracket distance. Similarly, space is frequently created within the arch as teeth are extracted to allow for alignment. Thus, the overestimation and model span utilized could be justified and serve as a foundation for the dentist to evaluate the archwire's force delivery [57].

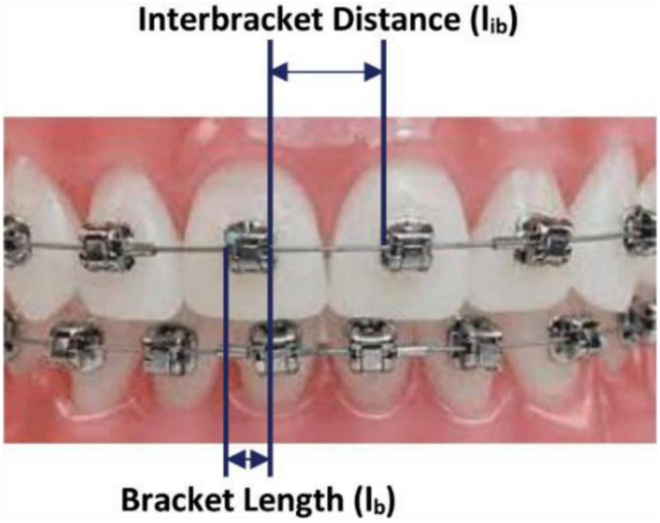

Fig. 4. Bracket length (l_b) and inter-bracket distance (l_{ib}) in MBT system labial brackets [56].

In prior study, it was noticed that the design of the bracket has a substantial impact on the amount of force released by superelastic NiTi alignment wires. The superelastic NiTi wires after 2 mm of maximum deflection are increased when 0.018-inch slot brackets are used instead of 0.022-inch slot brackets. After 4 mm of maximum deflection, the vertical slot size has no effect on the forces released by superelastic NiTi wires. They also found that, the use of a self-ligating bracket system increases the force generated by superelastic wire as compared to a traditional ligated bracket system [40]. In addition, it was found that the cross sections and dimensions of the wires also impact the force values, on average, higher in brackets with a 5 mm width than those with a 6.5 mm inter-bracket distance [58].

4. Effect of Bending Temperature

Macroscopic modifications develop on the surface of NiTi archwires due to the corrosive influence of the oral environment, bracket-wire interaction, or surface coatings [59]. The transition between a high-temperature austenite phase and a low-temperature martensite phase, nitinol wires have shape memory and superelasticity. Changes in the crystal lattice of the archwire material cause this transition, which can be caused by lowering the temperature or applying stress within a certain temperature range [60].

According to prior findings, all 0.019-0.025-inch NiTi wires showed distinct behaviors and were impacted differentially by temperature changes, signifying force delivery variance. Copper NiTi at 40 °C had the lowest constant force (200 g), followed by NeoSent. These wires would not function in mouth breathers since they're temperature sensitive [61]. It is worth noting that wires with a larger deflection generated a minimal deactivation force, resulting in a force difference. When an archwire recovered from a larger deflection, it generated a lower force than when it recovered from a smaller deflection. On the other hand, it was found that when the temperature was raised from 26 °C to 46 °C, the plateau slope rose from 0.66 N/mm to 1.1 N/mm [4]. Interestingly at temperatures of 35.5 °C and 44.0 °C, the superelastic NiTi wires exhibited a superelastic plateau of varied extent, which was seldom observed at 22 °C [62]. Clinically, temperatures of 5 °C and 20 °C are noteworthy because formability increases at these temperatures, making it simpler to shape the archwire before inserting it into the patient's mouth. Because austenite finish temperature (A_f) is above room temperature, a rise in temperature is required to create the superelastic effect. At human body temperature, all segments are completely austenitic [63].

5. Gingival/Labial Recession

It is vital to instill the importance of gingival recession in order to enhance the patient's face appearance during orthodontic therapy. Gingival recession (GR) is the exposure of the root surface caused by an apical tooth movement in the gingiva's location. The incidence and severity of gingival recession increases with age. Certain scientists have shown an association between gingival recession and mandibular incisor labial movement and have therefore identified this movement as a risk factor for gingival recession [64]. Four key areas are evaluated during the clinical assessment to prevent excessively aggressive orthodontic torques: the length of the incisor that creates the optimal position of the face gingival edge; the bone morphology beneath the gingiva; and the patient's gingiva thickness [65]. It has been found in earlier studies that rectangular archwires are appropriate for preoperative orthodontic preparation, severe torque movements should be avoided because of the increased risk of tooth loss and later GR [66].

6. Suitability of Archwire Force for Orthodontic Treatment

Archwires are critical in fixed orthodontic treatment because they generate the force required for tooth movement. They are capable of transmitting vertical (intrusion and extrusion) as well as horizontal (torque) forces. The effectiveness of these forces were determined by the wire's quality, kind and size [24]. These wires are directly measured in orthodontic force sources to obtain reliable measurement results. Much consideration was given to the link between the archwire's characteristics and the orthodontic force. The link between the bending characteristics of archwires and orthodontic force is erroneous. These studies cannot assist doctors in precisely estimating the orthodontic force generated by archwire design. In the past, researchers conducted orthodontic force measurement, statics research and finite element analysis to achieve accurate orthodontic force. The measurement and investigation of orthodontic force are generally carried out in the laboratory environment. The archwire is utilized to quantify the force, which avoids constricting the oral cavity's space and ensuring patient safety [67]. The correlation between the torque play, the bracket slot's height and the archwire's height and width expressed in Eq. (3) and shown in Fig. 5.

$$H = b.\sin\phi + h.\cos\phi \qquad (1)$$

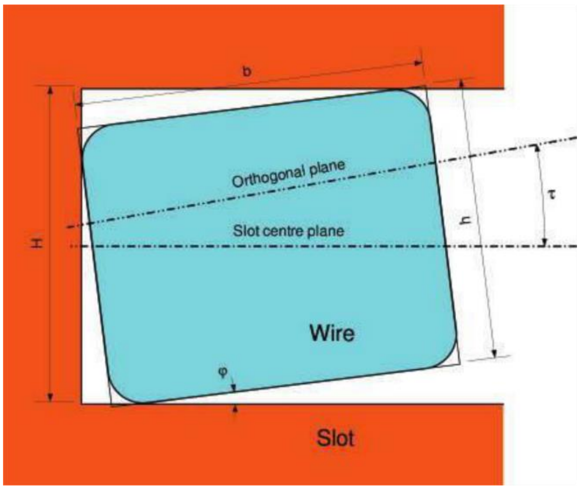

Fig. 5 Torque play between an archwire and the bracket slot [35].

Here H is the slot height, b and h are archwire width and height, and ϕ is the torque play [35]. In general, orthodontic treatment involves repositioning of teeth with the use of forces generated by appliances linked to the dentation. The orthodontic force, which has a direct impact on how effectively orthodontic therapy works, determines the pattern and rate of tooth movement. The orthodontic force following appliance loading is important for treatment planning, appliance design and optimization, and treatment prognosis. Orthodontic force can be measured theoretically, in vivo, and in vitro [44].

Table 1. Summary of bending setting vs. unloading force (Tooth movement).

Reference	Bending setting	Specifications	Unloading force (N)
[68]	Nitinol superelastic wire of 0.016 × 0.022 inch was deflected at the speed of 0.1 mm/min by maintaining 37 °C.	At deflection 3.0 mm	2.66
[69]	The round wire (0.014-inch) was chosen to generate unloading force deflection at 2.0 mm.	Self-ligated Elastomeric Elastomeric patterns	1.32 0.62 1.05
[58]	Superelastic NiTi 0.016 inch wires with 5 mm inter bracket width.	At deflection 2.0 mm At deflection 3.0 mm	3.40 4.33
[41]	0.016 inch NiTi archwires with self-ligation brackets were performed at a rate of 1 mm/min by maintaining 37 °C.	At deflection 3 mm	2.72
[1]	0.016 inch NiTi wires with a crosshead speed 0.01 mm/sec by maintaining 37 °C.	At deflection 2 mm	1.40
[40]	Superelastic 0.014 inch NiTi archwire with crosshead speed 0.01 mm/sec.	At deflection 3.5 mm (3PBT) At deflection 3.5 mm (SLB)	1.1 0.6
[16]	0.016-inch round NiTi subjected to three-point bending test.	At deflection 2.0 mm At deflection 4.0 mm	1.29 1.96
[44]	0.016 inch NiTi wires with a cross-head speed of 10 mm/min and placed in the brass vessel filled with artificial saliva maintained at 37 °C.	At deflection 1.5 mm At deflection 2.0 mm	1.64 2.93
[70]	Conventional NiTi wires with 0.016 inch with a loading velocity of 1 mm/min at 37 °C.	At deflection 2.0 mm	2.2
[24]	NiTi rectangular wires of 0.016 × 0.022 inch diameter. Wires were loaded to a maximum deflection from 0.5-1.5 mm using a 1 KN load cell at 5 mm/min in a water bath at 36±1 °C.	At deflection 0.50 mm At deflection 0.75 mm At deflection 1.00 mm At deflection 1.25 mm At deflection 1.50 mm	1.66 2.02 2.45 3.12 4.04
[54]	Superelastic NiTi 0.016 inch with low friction brackets, inter-bracket distance of 15 mm, crosshead speed of 7.5 mm/min at temperature 36±1 °C.	At deflection 3.0 mm	1.35
[71]	0.016 inch round thermal copper NiTi orthodontic wires were deflected 3 mm at 1 mm/min.	At deflection 3.0 mm	1.32
[52]	0.016-inch NiTi archwire with brackets 0.018-inch slot, at 0.05 mm/s and larger wire deflection at 2 mm and 4 mm at temperature 37 °C.	At deflection 2.0 mm At deflection 4.0 mm	1.80 2.86
[72]	0.016 × 0.022 inch NiTi with a length of 30 mm with a press speed of 5 mm/min.	At deflection 1.5 mm	3.93
[43]	NiTi archwire of 0.016 × 0.022 inch and 0.016 inch size with a length of 30 mm at temperature of 36 °C.	At deflection 3.0 mm (round) At deflection 3.0 mm (rect)	0.57-1.71 1.11-3.61

The findings of Table 1 evaluated on the basis of load/deflection graph with a flat slope upon unloading, known as the plateau, indicating that the force exerted is deactivation nearly constant across the range of tooth movement. This property is associated with the reversible transition from the austenitic to the martensitic phase over a particular stress threshold, which is met during activation [73]. The only parameter affected by the condition is the plateau slope. The plateau slope reflects an archwire's ability to exert more consistent stresses as displacement increases [74]. It is also evident that the magnitude of loading and unloading forces varies with respect to various parameters, such as archwire type, bracket size, and inter-bracket distance, and bending temperature but the most significant parameter is ageing treatment, which controls the superelasticity and shape memory effect of NiTi wires. In general, the maximum force highly depends on the shape of the wire, with rectangular wires always releasing higher forces. The maximum force decreases with increase inter bracket distance and the opposite trend was observed with wire deflection and testing temperature Moreover, the minimum force dropped correspondingly as the magnitude of the deflection increased. Consequently, a consistently lower minimum force was recorded for the group that was activated at a high deflection value. It is worth noting that this force range satisfies the force level requirement to achieve efficient tooth movement.

7. Conclusion

Orthodontics has benefited significantly over the last several decades from the development of NiTi wires and alloys. These wires offer a range of mechanical qualities, which promote orthodontic therapy and may boost patient comfort, minimize restorative dentistry time, and shorten treatment time. Superelasticity combined with the shape memory effect (SME) enables the orthodontist to accomplish more efficient tooth movement. The amount of loading and unloading forces vary with archwire type, bracket size, inter-bracket distance, and bending temperature, but ageing treatment influences the superelasticity and shape memory effect of NiTi wires. The ideal tooth force is 0.2 to 1.5 N. They can deliver outstanding treatment outcomes because they provide a low, constant force over a prolonged period of time, which is considered physiologically optimal for tooth movement.

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