

# An evaluation of the elastic recovery of nickel titanium archwires following ligation in palatally-displaced lateral incisors

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degree of Doctorate in Dental Surgery (D.Ch.Dent.) in Orthodontics

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## **Declaration**

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Leah Finan

## **Summary**

### **Introduction**

The levelling and aligning phase of orthodontic treatment involves the use of low-modulus nickel titanium archwires with small cross-sectional diameters.

### **Aims and objectives**

The aim of this study was to compare permanent deformation associated with a range of superelastic 0.012" and 0.014" nickel titanium archwires following ligation in a simulated set up. Furthermore, the study was designed to compare the maximum force imparted by these archwires at a variety of magnitudes of tooth displacement.

### **Materials and methods**

3D-printed models were created of a typodont with the lateral incisor palatally displaced (3-, 4- or 5- mm). The models incorporated conventional brackets, into which the archwires were ligated. A total of 288 archwires from two brands (Euroform (EO) and American Orthodontics (AO)) of two cross-sectional diameters (0.012" and 0.014") were tested for plastic deformation. This deformation was measured objectively using linear and angular measurements and subjectively using a validated index created to categorise archwires based on the degree of deformation and their clinical usability. 3–point bending tests were used to examine the load deflection characteristics of a sample of 72 archwires from the same 36 batches used in the typodont component of the study and determine the maximum and unloading force generated at archwire displacements of 3-, 4- and 5- mm.

## **Results**

Shapiro-Wilk normality tests showed the majority of the data collected for Part I was not normally distributed so therefore non-parametric Kruskal-Wallis testing was used. EF archwires had lower distortion compared to AO archwires ( $P = 0.008$ ). There was no statistically significant difference in deformation noted between 0.012" EF and AO archwires ( $P = 0.89$ ). However, the 0.014" EF archwire had significantly lower deformation than AO archwires ( $P = <0.001$ ). A wide variation in force levels at maximum archwire displacement was observed, influenced by archwire size and type, but with differences also noted within wire batches. Using independent samples Mann-Whitney U Tests, the mean force produced by the 0.012" archwires was significantly less than that produced by the 0.014" archwires at 3-, 4- and 5- mm displacements. The mean force produced by the AO archwires was significantly less than that for the EF archwires at the same magnitudes of displacement for both 0.012" and 0.014" sizes.

## **Conclusions**

Based on this artificial set-up, the use of an 0.012" archwire for tooth displacements of 3 mm or more may be considered in order to permit the delivery of lighter forces. EF archwires underwent a lesser degree of deformation than AO archwires. A high degree of variability in the performance of narrow dimension nickel titanium archwires can be expected and is influenced by manufacturer, archwire dimension and the degree of tooth displacement. Moreover, susceptibility to deformation appear to vary within batches of identical wires.

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## **Glossary of Terms**

| <b>Abbreviation</b> | <b>Description</b>                             |
|---------------------|------------------------------------------------|
| AO                  | American Orthodontics                          |
| EF                  | Euroform                                       |
| ISO                 | International Organisation for Standardisation |
| NiTi                | Nickel titanium                                |
| SIM                 | Stress-induced martensite                      |
| SS                  | Stainless steel                                |
| TTR                 | Temperature transition range                   |

# **Chapter 1 – Literature review**

## **1.1 Introduction**

Orthodontics is a specialised area of dentistry that focuses on the movement of teeth to correct malocclusions. The aim of orthodontic treatment is to achieve an aesthetic, functional and stable occlusion while maintaining an ideal facial balance (Proffit, 2019). In 1972, Andrews through his six keys described the ideal occlusion (Andrews, 1972). In order to achieve the ideal occlusion, tooth movements are generally carried out using fixed appliances consisting of archwires which engage into brackets attached to the individual teeth. This in turn transmits a force to the dentition resulting in tooth movement (Kusy, 1997).

There is no one ideal archwire, with stainless steel, nickel titanium, beta-titanium and cobalt-chromium being in widest use. Each alloy has properties which make them more or less suited to the various stages of orthodontic treatment. The initial stage of orthodontic treatment is known as the levelling and alignment phase. At the beginning of this stage, the teeth are typically malaligned and in crowded positions within the dental arch. A flexible archwire with a small cross-sectional diameter is required to allow engagement of the archwire into the bracket slot. Typically, nickel titanium is used for this stage of treatment as it possesses suitable properties such as high springback and has a low modulus of elasticity (Nikolai, 1997; Proffit, 2019). These properties allow the archwire to be engaged in the bracket slots of malaligned teeth and deliver low continuous forces to the teeth over

prolonged periods thus allowing controlled tooth movement (Nikolai, 1997; Proffit, 2019).

Permanent deformation of initial archwires may occur, particularly in arches with more displaced teeth and due to individual patient factors including higher muscular forces which increase the likelihood of distortion. These factors can result in an archwire being no longer effective in delivering forces to the teeth, necessitating replacement and reactivation, risking extension of treatment duration.

## **1.2 Tooth movement**

Orthodontic tooth movement is a process of applying external forces to a tooth which initiates a number of responses (Meikle, 2006). If prolonged light pressure is applied to a tooth, the bone around the tooth will remodel and tooth movement will occur (Proffit, 2019).

Ideally, light continuous forces should be used in order to achieve safe and efficient orthodontic tooth movement (Meeran, 2012). Optimal orthodontic force magnitude has been defined as the amount of force that can allow a maximal rate of orthodontic tooth movement with minimal tissue damage and patient discomfort (Proffit, 2019). This 'light' force is considered an arbitrary term as there is no scientific evidence currently available on the ideal force for orthodontic tooth movement (Ren *et al.*, 2003). However, it is accepted that heavier forces result in occlusion of the blood vessels leading to hyalinisation, necrosis and undermining resorption (Krishnan & Davidovitch, 2006).

Two types of tooth movement can occur based on the magnitude of the force delivered to the tooth:

### 1. Frontal resorption:

During this process, whereby a light sustained force is applied to a tooth, the periodontal ligament becomes compressed but the blood vessels and the capillaries remain patent. This stimulates cellular activity allowing fibroblasts to proliferate and osteoclasts migrate from the capillaries to the periodontal ligament (Gianelly & Schnur, 1969). Osteoclasts resorb the lamina dura from the area adjacent to the compressed part of the periodontal ligament which allows tooth movement to occur. It takes approximately 2 days for tooth movement to commence (Proffit, 2019).

### 2. Undermining resorption:

This process occurs when the sustained force delivered to the tooth is large enough to occlude the blood vessels of the periodontal ligament. Sterile necrosis of all the cells within the compressed part of the periodontal ligament will occur. This process has been described as hyalinisation due to the histological appearance of the PDL whereby all of the cells have disappeared (Reitan, 1951).

Remodelling of the bone must then be completed by cells recruited from the undamaged area of the periodontal ligament adjacent to the hyalinised zone. This cell recruitment into the hyalinised zone results in a delay of a number of days. Osteoclasts start to appear in the adjacent bone marrow spaces and begin resorbing from the underside of the lamina dura. This process has been described as undermining resorption (Rygh, 1974). This results in a further delay as a substantial amount of bone must be removed before any tooth

movement can occur. Tooth movement by this process usually requires 7 to 14 days (Proffit, 2019).

### **1.3 Levelling and alignment stage of orthodontic treatment**

The levelling and alignment phase of orthodontic treatment is considered the first stage of treatment and deals mostly with the correction of dental crowding and rotations. An ideal archwire for this phase of treatment will have high strength, low stiffness and a long working range (O'Brien *et al*, 1990).

This phase is characterised by using archwires which progressively increase in diameter as the teeth align. Nickel titanium alloys have favourable mechanical properties for this stage given their low modulus of elasticity and low deflection load. This allows the application of a force of long duration and low magnitude allowing biologically effective tooth movement.

(Wang *et al.*, 2010).

### **1.4 Efficiency of orthodontic alignment**

Previous studies have attempted to ascertain whether nickel titanium archwires are superior to other materials in terms of alignment efficiency, without consensus (Evans *et al.*, 1998; O'Brien *et al.*, 1990). Wang et al. (2018) conducted a Cochrane review of the effect initial archwires used in orthodontic treatment have on the rate of tooth alignment, the degree of root resorption and the level of pain experienced by the patients. A total of 12 randomised controlled trials were included in the review and 6 comparisons

were assessed as follows: Multistrand stainless steel versus superelastic nickel titanium archwires, multistrand stainless steel versus thermoelastic nickel titanium archwires, conventional nickel titanium versus superelastic nickel titanium archwires, conventional nickel titanium versus thermoelastic nickel titanium archwires, single-strand superelastic nickel titanium versus coaxial superelastic nickel titanium archwires and superelastic nickel titanium versus thermoelastic nickel titanium archwires. In terms of results, moderate-quality evidence showed that coaxial superelastic nickel titanium archwires can produce greater tooth movement over a 12-week period than single-strand superelastic nickel titanium archwires. Moderate-quality evidence also suggests that at day 1 there is no difference in pain levels between multistrand stainless steel arch wires and superelastic nickel titanium archwires. This review deemed insufficient evidence exists to determine if any particular archwire material is better than any other in terms of alignment rate, time taken to align the teeth, pain or root resorption.

In 2022, Wazwaz *et al.* carried out a systematic review, including 35 trials, looking at the duration of tooth alignment of the mandibular dentition with fixed appliances. Whole arch alignment of the mandibular dentition was found to be 263 days while mandibular incisor alignment was found to be 100 days. The review concluded that patient- and treatment- related characteristics can significantly affect the duration. However, there was no significant difference in incisor alignment duration between different nickel titanium archwires.

Some of the studies comparing the alignment efficiency of different initial aligning archwires included in the Cochrane and systematic review are outlined in Table 1.1. Abdelrahman *et al.* (2015), Atik *et al.* (2019) and Azizi *et al.* (2021) compared tooth alignment while using different types of nickel titanium archwires, namely, conventional nickel titanium, superelastic nickel titanium and heat-activated nickel titanium. No statistically significant difference in tooth alignment efficiency was noted. Sebastian *et al.* (2012) compared different configurations of nickel titanium archwires, including coaxial versus single stranded archwires. This study concluded that there was a statistically significant difference in the alignment of the lower labial segment favouring the coaxial configuration.

When comparing alignment efficiency using different archwire materials, Quintao *et al.* (2005) compared superelastic nickel titanium with stainless archwires and found no statistically significant difference in tooth alignment efficiency in both the maxillary and mandibular arches. Nordstrom *et al.* (2018) also compared archwires of differing materials and concluded there was no difference in aligning efficiency of the lower labial segment when using nickel titanium versus niobium-titanium-tantalum-zirconium archwires.

Mandall *et al.* (2006) and Ong *et al.* (2011) compared alignment efficiency while using three different archwire sequences. No significant difference was noted in either study in relation to the time taken to reach a working archwire.

**Table 1.1. Initial archwire type and efficiency of alignment**

| <b>Author /<br/>Year /<br/>Study Type</b>                               | <b>Publication</b>                                                                                                                                                                                                                                                                     | <b>Intervention</b>                                                          | <b>Results</b>                                                                                                                     |
|-------------------------------------------------------------------------|----------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|------------------------------------------------------------------------------|------------------------------------------------------------------------------------------------------------------------------------|
| Abdelrahman <i>et al.</i><br>(2015)<br><br>RCT involving 74<br>patients | A clinical<br>comparison of<br>three aligning<br>archwires in<br>terms of<br>alignment<br>efficiency: A<br>prospective<br>clinical trial                                                                                                                                               | Conventional<br>NiTi<br><br>v superelastic<br>NiTi v heat-<br>activated NiTi | No differences<br>in efficiency of<br>aligning lower<br>arch at 8 and<br>16 weeks<br><br>(P = 0.98)                                |
| Atik <i>et al.</i> (2019)<br><br>RCT involving 50<br>patients           | A comparative<br>assessment of<br>clinical<br>efficiency<br>between<br>premium heat-<br>activated<br>copper nickel-<br>titanium and<br>superelastic<br>nickel-titanium<br>archwires<br>during initial<br>orthodontic<br>alignment in<br>adolescents: a<br>randomized<br>clinical trial | Superelastic<br>NiTi v heat-<br>activated NiTi                               | No statistically<br>significant<br>difference<br>between<br>groups in<br>terms of<br>efficiency of<br>alignment<br><br>(P = 0.581) |

|                                                                        |                                                                                                                                                                                                          |                                                |                                                                                                                                                                                                                                                               |
|------------------------------------------------------------------------|----------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|------------------------------------------------|---------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
| <p>Azizi <i>et al.</i> (2021)</p> <p>RCT involving 88 patients</p>     | <p>Tooth alignment and pain experience with A-NiTi versus Cu-NiTi: a randomized clinical trial</p>                                                                                                       | <p>Superelastic NiTi v heat activated NiTi</p> | <p>No statistically significant difference between groups in terms of efficiency of alignment of the lower labial segment (P &gt; 0.05)</p>                                                                                                                   |
| <p>Sebastian <i>et al.</i> (2012)</p> <p>RCT involving 24 patients</p> | <p>Alignment efficiency of superelastic coaxial nickel-titanium vs superelastic single-stranded nickel-titanium in relieving mandibular anterior crowding: a randomized controlled prospective study</p> | <p>Coaxial NiTi v round NiTi</p>               | <p>A statistically significant difference (P &lt; .05) between groups in terms of efficiency of alignment of the lower labial segment favouring the coaxial superelastic NiTi over single-stranded superelastic NiTi in relieving lower anterior crowding</p> |

|                                                                        |                                                                                                                                                                                                                                        |                                                                                                  |                                                                                                                                                                               |
|------------------------------------------------------------------------|----------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|--------------------------------------------------------------------------------------------------|-------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
| <p>Quintao <i>et al.</i> (2005)</p> <p>RCT involving 45 patients</p>   | <p>A prospective clinical trial to compare the performance of four initial orthodontic archwires</p>                                                                                                                                   | <p>Superelastic NiTi v heat activated NiTi v stainless steel v multistranded stainless steel</p> | <p>No significant difference between the different archwires groups in terms of efficiency of alignment of the lower labial segment as well as upper and lower whole arch</p> |
| <p>Nordstrom <i>et al.</i> (2018)</p> <p>RCT involving 28 patients</p> | <p>Comparison of changes in irregularity and transverse width with nickel-titanium and niobium-titanium-tantalum-zirconium archwires during initial orthodontic alignment in adolescents: A double-blind randomized clinical trial</p> | <p>NiTi v niobium-titanium-tantalum-zirconium archwires</p>                                      | <p>No difference between groups in terms of efficiency of alignment (Little's irregularity index) (P = 0.29)</p>                                                              |

|                                                                       |                                                                                                                 |                                                                                                                                                                                                                                |                                                                                                                                                                                                                                                                                           |
|-----------------------------------------------------------------------|-----------------------------------------------------------------------------------------------------------------|--------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|-------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
| <p>Ong <i>et al.</i> (2011)</p> <p>RCT involving 132 patients</p>     | <p>Alignment efficiency and discomfort of three orthodontic archwire sequences: a randomised clinical trial</p> | <p>Round NiTi v rectangular heat activated NiTi v Sentalloy</p>                                                                                                                                                                | <p>No difference between groups in terms of efficiency of alignment of the lower labial segment or the time taken to reach working archwire (P = 0.28)</p>                                                                                                                                |
| <p>Mandall <i>et al.</i> (2006)</p> <p>RCT involving 150 patients</p> | <p>Which orthodontic archwire sequence? A randomized clinical trial</p>                                         | <p>3 groups:</p> <p>A = 0.016" NiTi, 0.018 x 0.025" NiTi, 0.019 x 0.025" SS</p> <p>B = 0.016" NiTi, 0.016" SS, 0.020" SS, and 0.019 x 0.025" SS</p> <p>C = 0.016 x 0.022" CuNiTi, 0.019 x 0.025" CuNiTi, 0.019 x 0.025" SS</p> | <p>No significant difference between groups for patient discomfort (P &gt; 0.05), root resorption (P = 0.58) or time taken to reach working archwire (P &gt; 0.05). Statistically significant difference in number of visits to working archwire for group B than group A (P = 0.012)</p> |

## **1.5 Orthodontic archwires**

Orthodontic archwires can be classified in many ways based on material, cross-sectional shape, size or archform type.

### **1.5.1 Archwire materials**

#### **1.5.1.1 Gold alloys**

In the early 1900s archwires derived from gold alloy were commonly used by orthodontists as they were considered stable enough to be used in the oral cavity. However they are no longer used due to their low yield strength, difficulty in manipulation, lack of springback properties and their increasing cost (Proffit, 2019).

#### **1.5.1.2 Stainless steel**

Austenitic stainless steel became popular in the 1940s replacing gold alloys in the manufacturing of orthodontic archwires. The most common composition of the stainless steel alloys used contain approximately 72% iron, 18% chromium, 8% nickel and 0.2% carbon (Proffit, 2019). Stainless steel archwires are relatively stiff and are typically used as a working archwire once the levelling and aligning phase is complete. Historically, however, these archwires were used in the levelling and aligning phase of treatment. The flexibility of the archwire was increased by increasing the length of the archwire through the addition of loops. This allowed engagement into the bracket slots of the

malaligned teeth but was a time-consuming process. Multistranded stainless steel archwires can be used in the alignment phase of orthodontic treatment. These archwires are manufactured by twisting two or more strands of small diameter wire (<0.01 inch) into an multistranded cable. The flexibility of these archwires is influenced by how tightly woven the cable is as well as the characteristics of the individual strands of wire (Evans *et al.*, 1998).

#### **1.5.1.3 Cobalt - Chromium (Elgiloy)**

Cobalt-Chromium is an alloy with a typical composition of 40% cobalt, 20% chromium, 15% nickel, 15.81% iron, 7% molybdenum, 2% manganese, 0.15% carbon and 0.04% beryllium. This alloy was originally developed in the 1950s by the Elgin Watch Company to be used as a watch spring. Rocky Mountain Orthodontics then marketed it as Elgiloy for orthodontic archwires. There are four tempers of this alloy available for orthodontic use.

#### **1.5.1.4 Nickel titanium**

Nickel titanium alloys are considered the material of choice for the initial alignment phase due to their ability to deliver a light force over a large range of activation. These alloys are capable of existing in two crystalline forms, martensitic and austenitic, with different atomic arrangements which give rise to their main properties (Brantley, 2001; Kusy, 1997). The words martensite and austenite originate from the two prominent metallurgists, Adolf Martens and Sir William Chandler Roberts-Austen.

The martensitic form has a body-centred, cubic or tetragonal, crystal structure and generally exists and is more stable at low temperatures and high stress (Figure 1.1). The austenitic form has a face-centred, hexagonal close-packed, crystalline structure and generally exists and is more stable at higher temperature and lower stress (Brantley, 2001). The presence of the austenitic phase results in a more rigid alloy as the archwire will show a maximum of 2% elastic deformation within the strain limit. In the martensitic phase, the archwire is capable of elastic deformation of 8 – 10% before permanent plastic deformation occurs (Otsuka & Wayman, 1999; Santoro *et al.*, 2001a).

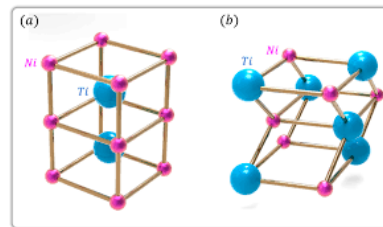


**Figure 1.1. Austenitic vs martensitic forms (Shimoga, Kim, & Kim, 2021)**

Phase transformation between these forms can occur. An intermediate phase, known as the R phase, can form between the austenitic and martensitic phase. This forms during the cooling or heating transformations. The crystalline structure of this phase is a simple hexagonal lattice with rhombohedral symmetry (Brantley, 2001). The phase transformations which occur between the austenitic and martensitic forms are fully reversible which makes nickel titanium alloys unique. Phase transformation occurring between these two phases can be measured using radiographic diffraction looking at the different crystal structures. Temperature-modulated differential scanning calorimetry can also be used to provide information on the transformations between the phases (Bradley *et al.*, 1996; Brantley *et al.*, 2003). The phase transformation

between the austenitic and martensitic forms give rise to the properties of superelasticity and shape memory which are absent in other archwire materials (Proffit, 2019). These properties are particularly useful in the initial alignment stage of orthodontic treatment. The terms superelasticity and pseudoelasticity are often used in the literature interchangeably when discussing nickel titanium. It is a qualitative mechanical property and can be broadly defined by the stress-strain plot (Figure 1.2). It occurs when the stress value remains fairly constant up to a point of wire deformation. It is characterised by a load-deflection plot with a horizontal region during loading which implies that a constant force is exerted over that particular range of tooth movement. These archwires supposedly exert the same force irrespective of whether they are deflected at a small or large distance (Evans *et al.*, 1998). During unloading, the archwire reverts from a martensitic phase to an austenitic phase while imparting a low continuous force onto the teeth as they align (Burstone *et al.*, 1985; Miura *et al.*, 1986; Thayer *et al.*, 1995). Shape memory is the ability of an alloy to return to its original shape after undergoing plastic deformation in the martensitic phase (Kusy, 1981). In a non-shape-memory alloy, the strain of deformation is absorbed by the rearrangement of crystals and the crystals are unable to return to their original position (Hurst *et al.*, 1990). With a shape memory alloy, changes occur at an atomic level whereby the crystals will stay in their original place but the atoms within the crystals rearrange. This allows the distorted metal to revert to its original form and there will be no visible changes to the alloy (Miura *et al.*, 1986).

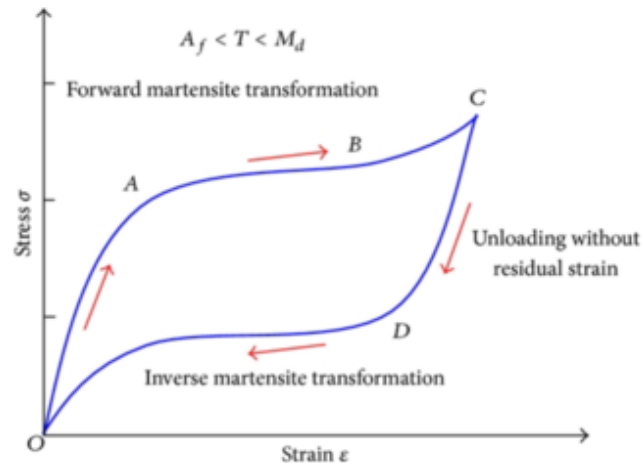


**Figure 1.2. Stress-strain graph representing a superelastic nickel titanium archwire. The unused archwire has an austenitic structure. As the archwire is placed under stress, through engagement into the bracket of a displaced tooth, it initially demonstrates linear elastic behaviour (to point A on the graph). As the archwire is ligated into the bracket, beyond 2mm of displacement, stress-induced martensitic transformation occurs within the wire allowing the archwire to become more pliable in its martensitic phase. This is represented by the loading plateau from point A to B showing that the stress within the wire remains relatively constant as the archwire is further deflected, the superelastic effect. Points B to C represent martensitic transformation throughout the archwire and the archwire beginning to demonstrate a linear pattern once again. Once the teeth start to move, there is a sharp reduction in the stress within the archwire due to hysteresis as well as friction within the bracket/archwire system (point C to D). The stress within the archwire will then begin to plateau once again as the alloy transforms back to its austenitic phase. Clinically, low forces are imparted to the tooth over a long range of action while the archwire returns to its original shape, demonstrating shape memory (Qian, Li, Song, & Guo, 2013).**

Heat activation refers to the ability of nickel titanium alloys to go from the martensitic to the austenitic phase as its temperature rises. This heat-activated change in the crystalline structure is known as thermoelasticity. This ability of nickel titanium alloys gives rise to its shape memory properties. Nickel titanium alloys can be deformed plastically in the martensitic phase but will then return to their original shape and their austenitic phase if they are heated through a certain transition temperature range (Miura *et al.*, 1986).

Shape memory is also known as thermoelasticity and refers to the temperature-induced change in the crystalline structure (Fernandes, 2001). At low temperatures the alloy is soft and is capable of changing shape. As the temperature rises to a suitable temperature transition range, the alloy can be reformed into its original configuration. This phenomenon is known as heat-induced martensitic transformation (Miura *et al.*, 1986). Shape memory properties can be altered by adding a third element e.g. through the addition of copper. These properties can be confirmed through a stress-strain curve with plateaus which do not exist with martensitic stabilised alloys. The difference between these plateaus accounts for the loss of energy due to crystalline structure alteration, known as hysteresis (Figure 1.2). The temperature at which the parent high-temperature austenitic form begins to transform into the low-temperature martensitic form is known as the transformation temperature. There are three broad classes of commercially available nickel titanium arch wires with different stress-strain relationships: (i) martensitic stable (ii) austenitic active and (iii) martensitic active (Kusy, 1997).

#### **1.5.1.4.1 Martensitic stable e.g. Nitinol**

Nickel titanium was originally developed by the Naval Ordnance Laboratory in the United States in the 1960s through the work of Buehler. This was called Nitinol based on an acronym for Nickel Titanium Naval Ordnance Laboratory (Buehler, 1963). Interest developed in this alloy for medical use due to its high elastic limit and low modulus of elasticity (Miura, 1986). In the early 1970s, Dr George Andreasen and his team reported on his clinical studies using nickel titanium archwires and Unitek subsequently introduced Nitinol to orthodontists (Andreasen & Hilleman, 1971; Miura *et al.*, 1988).

This initial nickel titanium archwire is classified as martensitic stable with its crystalline structure being fixed in the martensitic form and has a 50:50 composition of nickel and titanium. It has good springback properties and a low modulus of elasticity but, as a consequence of being manufactured through a work hardening process, it does not have any shape memory or superelastic properties. Cold working by more than 8-10% suppressed the shape memory effect (Andreasen & Morrow, 1978; Kusy, 2002; Miura *et al.*, 1988). Nevertheless, this alloy has attractive properties including an increased working range, allowing the archwire to be deflected into the brackets and is capable of delivering light forces to the teeth of approx. one-fifth of the force per unit activation of other available alloys at the time (Kusy, 1997).

Martensitic stable alloys are flexible which makes them useful when large archwire deflections are required and show linear behaviour on a stress-strain diagram (Ferreira, 2010). Clinically, these archwires can be deformed

permanently when they exceed a certain limit or if they are left in the mouth for lengthy periods exposed to intra-oral cycling (Ferreira, 2010).

#### **1.5.1.4.2 Austenitic active**

Superelastic nickel titanium archwires are classified as austenitic active. These archwires have primarily an austenitic crystalline structure and may have some martensitic nickel titanium also present. They have a temperature transition range (TTR) below that of the oral cavity and therefore undergo martensitic transformation by mechanical deformation. As the archwire is deflected beyond a certain point ( $>2$  mm or  $50^{\circ}$  -  $70^{\circ}$ ), stress-induced martensite (SIM) occurs in localised areas of the wire (Evans *et al.*, 1998; Santoro *et al.*, 2001b). Clinically, this allows the archwire to engage in the bracket slots of malaligned teeth as the archwire is more pliable in the martensitic state. The superelastic property is represented by a plateau on the load deflection curve, whereby the archwire is capable of delivering the same force over a large range of deflection. This archwire will remain superelastic until the teeth begin to align. As the archwire deflection reduces with tooth alignment, the stiffer austenitic phase will return (Santoro *et al.*, 2001a). Due to the manufacturing challenges incurred while making austenitic archwires there can be a difference in their elastic properties between different manufacturers as well as within the same manufacturers (Bourauel *et al.*, 1997).

#### **1.5.1.4.3 Martensitic Active**

These archwires can be referred to as thermally active, heat-activated or thermoelastic nickel titanium wires e.g. Sentalloy (Kusy, 1997). Shape memory is the ability of an archwire to remember its form and return to it through the application of stress or a change in temperature (Miura *et al.*, 1986). Only archwires in the austenitic phase are capable of memorising the pre-determined shape. The shape is usually determined at higher temperatures and these archwires are said to have a thermally-induced shape memory effect (Kusy, 1997). Through the addition of copper to the alloy, the austenitic transformation temperature can be better controlled. However, the addition of chromium is necessary to counteract the rise in transformation temperature above body temperature which occurs with the addition of copper (Melton, 1990). Each alloy has a temperature transition range during which a stress-induced phase transformation of the crystalline structure occurs (Parvizi & Rock, 2003). Therefore clinically, these archwires are predominantly in the martensitic phase at room temperature meaning that they are flexible as they are placed into the brackets of displaced teeth within the oral cavity. The archwire will increase in temperature to that of the oral cavity. Typically, these archwires are designed to have a temperature transition range similar to the temperature of the oral cavity (approximately 35 °C) (Miura *et al.*, 1986; Parvizi & Rock, 2003). When the archwire is heated to beyond this range, stress-induced phase transformation will occur. As the archwire returns to its austenitic phase, it will 'remember' its original shape and will place a force on the teeth producing orthodontic tooth movement (Kapila & Sachdeva, 1989).

There are four phase transformation temperatures typically specified for nickel titanium orthodontic archwires. The austenite start temperature ( $A_s$ ) is the start of the transformation from martensite to austenite when the wire is heated. The austenite finish temperature ( $A_f$ ) is the end of the transformation from martensite to austenite when the wire is heated. The martensitic start temperature ( $M_s$ ) occurs at the start of the transformation from austenite to martensite when the wire is cooled while the martensitic finish ( $M_f$ ) temperature is at the end of the transformation from austenite to martensite when the wire is cooled (Brantley, 2001). The ISO standard states that the  $A_f$  temperature for type 2 wires, determined by differential scanning calorimetry, must be specified by each company to the nearest degree Celsius (ISO, 2014).

Various classification systems for nickel titanium alloys have been developed over time. In 1992, Waters developed a classification system for nickel titanium archwires based on their temperature transition ranges:

Group 1: Alloys with TTR between room temperature and body temperature (martensitic active alloys)

Group 2: Alloys with TTR below room temperature (austenitic)

Group 3: Alloys with TTR close to body temperature – these will spring back to their original shape when activated by body heat (Waters, 1992).

Evans and Durning (1996) developed a more comprehensive classification of alloys used in orthodontic treatment:

Phase I: alloys such as gold and stainless steel

Phase II: stabilised, referring to original work hardened Nitinol

Phase III: superelastic-active austenite, predominantly austenitic at room temperature with TTR below the intraoral temperature and therefore presents with pseudoelastic properties.

Phase IV: thermodynamic or active martensitic. Due to improvements in the ability to alter the amount of austenite present in the alloys, these alloys have a TTR set very close to the intra-oral temperature.

Phase V: graded thermodynamic

#### **1.5.1.5 Titanium molybdenum alloy (Beta-titanium)**

In 1979, titanium molybdenum alloy archwires were introduced by Goldberg and Burstone for orthodontic use. It is an alloy composed of 79% titanium, 11% molybdenum, 6% zirconium and 4% tin (Goldberg & Burstone, 1979). These archwires have properties between stainless steel and nickel titanium and have some favourable properties such as a low modulus of elasticity, good spring back properties and are formable. They have one third of the stiffness of stainless steel archwires and are twice as stiff as martensitic stabilised nickel titanium (Krishnan & Kumar, 2004). Their main drawbacks are their high surface roughness which results in an increase in friction between the bracket and the archwire during sliding mechanics. Reduction of the friction of these archwires has been attempted using ion implantation with nitrogen ions. However, Kula *et al.* (1998), carried out a split mouth study which found no difference in the rate of space closure between the treated versus untreated titanium molybdenum alloy archwires.

## **1.5.2 Archwire properties**

### **1.5.2.1 Physical features**

The physical features of orthodontic archwires can be expressed using a stress-strain or load-deflection diagram. Orthodontic archwires tend to have elastic behaviour characteristics meaning that, as the archwire is deflected, it will impart a force in response which is dependent on the rigidity of the archwire. A more vertical slope will be displayed if the archwire is rigid while a more horizontal slope represents a more flexible archwire.

The resilience of an archwire is represented by the area under the load-deflection curve up to the proportional limit and represents the energy storage capacity of the archwire. The larger the proportional limit, the more resilient the archwire is. The proportional limit is defined as the point at which permanent deformation starts to occur. Clinically, the higher proportional limit means that the wire can be deflected to greater degrees before permanent deformation will occur. It is desirable for an orthodontic archwire to have high resilience in order for it to elicit the stored energy it has absorbed while it was deflected to align the teeth (Figure 1.3).

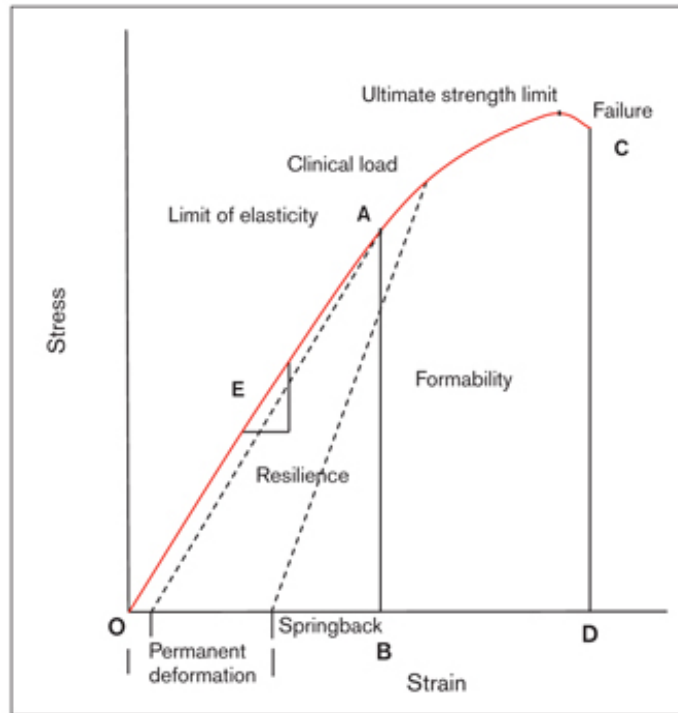


**Figure 1.3. Typical stress-strain diagram of an orthodontic archwire (Proffit, 2019)**

#### **1.5.2.2 Ideal archwires**

An ideal archwire will have the following properties to allow safe and efficient tooth movement:

- Ability to deliver a constant force over a wide arc of deflection
- Produce clinically acceptable levels of biomechanical forces when activated in the oral cavity
- Have adequate ability for clinical manipulation
- High strength to withstand deformation
- High recoverable elastic strain to increase the wire working time
- High resilience to possess energy for tooth movement

- Formable so that the wire can be bent into shape
- Joinable through welding / soldering for certain clinical applications
- Biocompatible and environmentally stable
- Be friction free
- Be resistant to corrosion in oral environment
- Have a low cost

However, there is currently no specific archwire available, for use in orthodontics, which possesses all of these desirable properties. A combination of archwire types and cross-sectional sizes are generally required during a patients orthodontic treatment. Clinical selection depends on the case and clinical situation, and the combination of properties offered by each wire (Kusy, 1997).

#### **1.5.2.3 Initial levelling and aligning archwires**

In the initial stage of orthodontic treatment the aim is to level and align the dentition. At the beginning of this stage of treatment, the teeth are generally in their most malaligned and crowded positions. In order for alignment to take place an archwire must engage in the brackets attached to the teeth. In these crowded arches. There are certain desirable characteristics that an archwire should possess to allow for this as follows:

- Good springback properties which allows the archwire to be deflected over a large distance without being permanently deformed. This allows the archwire to be deflected into crowded and malaligned dental arches without permanent deformation. This allows tooth movement to

be controlled and allows for greater intervals between activations (Burstone *et al.*, 1985).

- Good ability to exert light and constant forces capable of producing an adequate biological response in the periodontal ligament to achieve physiological bone remodelling to allow tooth movement (Proffit and Fields, 1993).

Nickel titanium is one such alloy possessing properties suited to initial alignment.

### **1.5.3 Manufacturing process of orthodontic archwires**

It took almost a decade for nickel titanium to become commercially available after it was initially discovered. This was due to the complex manufacturing process compounded by the difficulty in machining the wire. Nickel titanium is composed of 50-51% nickel by atomic weight and 55-56% by weight percent (Otsuka & Wayman, 1999). Small changes to the relative percentages of nickel and titanium have large effects on the transformation temperatures of the alloy. Vacuum induction melting, electron beam melting, plasma melting or vacuum arc melting processes are used in the creation of nickel titanium alloys (Wang *et al.*, 2010). The alloy goes through multiple reheats to improve the uniformity of the alloy. This is then converted into a powdered form. Isostatic processing is then used to make wires from the powdered alloy. The alloy is hot worked followed by a final heat treatment and then it is formed into the final shape by drawing and rolling processes. A drawing process is used in the manufacturing of round archwires. This involves the original cast ingot of the

alloy being subjected to various stages of intermediate heat treatments (Brantley, 2001). A Turks Head machine is used to create rectangular and square archwires. During this manufacturing process, a round archwire is rolled through the machine which has vertical and horizontal rollers incorporated to shape it.

#### **1.5.4 Manufacturers of orthodontic archwires**

There are a number of companies that supply orthodontic archwires. Some of these companies manufacture and supply their own archwires while others market and supply archwires manufactured by others. The marketing strategies of some of these companies is to promote certain brands as being superior in terms of their properties and clinical performance. However, not only is the exact composition of nickel titanium archwires i.e. the ratio of each metal often withheld by the manufacturers but very often their properties are also unavailable to the consumer. Additionally, variations in measuring conditions and specifications between manufacturers make it difficult to compare archwires (Wilkinson *et al.*, 2002).

#### **1.6 Laboratory testing of archwires**

The mechanical properties of archwires are generally tested using tensile, bending, deflection and torsional laboratory-based tests. These tests offer reproducibility and allow the properties of archwires to be compared with each other but their main limitation is that they do not reflect variables

encountered in clinical situations (Bartzela *et al.*, 2007; Wilkinson *et al.*, 2002). In 2006, the American National Standard/American Dental Association (ANSI/ADA) Specification No 32 for orthodontic wires was developed by a committee of members from universities, orthodontic companies and private practices. This stipulated a cantilever bending test to evaluate mechanical properties of orthodontic archwires. Mechanical property evaluation of orthodontic archwires should follow the current International Organisation for Standardisation (ISO) standard which has been adopted by the American Dental Association. ISO 15841:2014 specifies the recommended requirements and methods to be used in the evaluation of wires inherent in fixed and removable orthodontic appliances as well as giving detailed requirements concerning the presentation of the physical and mechanical properties of orthodontic wires, the test methods by which they can be determined, and packaging and labelling information.

According to the ISO standard, orthodontic archwires are classified as Type 1 if they display linear elastic behaviour (including stainless steel, cobalt-chromium and beta-titanium wires) or Type 2 if they display non-elastic behaviour (nickel titanium) during unloading at temperatures up to 50 °C. Type 1 archwires can be tested in tension or three-point bending and must conform to the testing conditions and specimen dimensions provided in the ISO standard. Type 2 archwires can be tested using three-point bending tests at  $36\text{ °C} \pm 1\text{ °C}$  only. Nickel titanium archwires are classified as Type 2 archwires and therefore should be tested using three-point bending tests using a 10 mm span of archwire within a temperature range of  $36\text{ °C} \pm 1\text{ °C}$ . During the test,

six archwires from each batch are deflected to a distance of 3.1 mm and the force is recorded at deflections of 0.5 mm, 1 mm, 2 mm and 3 mm (ISO, 2014).

### **1.6.1 Testing load deflection properties**

Load deflection properties are typically evaluated using cantilever-type 3-point wire bending test. These tests were first described by Miura in 1986 when examining the difference between the conventional nitinol archwires and superelastic archwires (Miura *et al.*, 1986). Three-point wire bending tests are used to determine the activation and deactivation forces of a particular archwire. The activation force or loading force of an archwire describes the force required to engage the archwire into the bracket slot while the deactivation force or unloading force describes the force applied by the archwire to the teeth in returning to its original shape (Bhat, 2018). This is represented by a load deflection diagram with the upper curve being the loading force while the lower curve represents the unloading force (Figure 1.2).

There is currently no established consensus regarding the exact method through which three-point bending tests should be completed (Elayyan *et al.*, 2010). However, it is considered the gold standard method of testing the mechanical properties of an archwire and offers reproducibility allowing comparison between studies (Bartzela *et al.*, 2007; Kapila & Sachdeva, 1989). One limitation is the inability of current clinical trials to determine or confirm that superelastic nickel titanium archwires perform better clinically than nitinol or multistranded stainless-steel archwires during

initial alignment (Evans *et al.*, 1998). To overcome these limitations and to better represent clinical scenarios, it has been suggested that testing is carried out with the archwire constrained as part of a fixed appliance (Waters *et al.*, 1975). It has been noted in the literature that the design of the testing models can influence the load deflection plots of archwires which must be noted when comparing archwires (Oltjen *et al.*, 1997; Rock & Wilson, 1988). Oltjen *et al.* (1997) compared traditional 3-point bending tests with 3-bracket tests and found a more constrained mode of bending when brackets were used due to the increase in friction from ligation as well as a degree of binding at the bracket-wire interface.

Cantilever testing has been used also as a method of testing the mechanical properties of orthodontic archwires. Ideal testing methods should simulate clinical orthodontic scenarios as closely as possible. This is not possible with unilateral beam testing based on the free-end beam principle whereby the archwire is unsupported at one end which being deflected on the other. Friction is introduced as the archwire slides over the support during testing which results in an increased loading force which can simulate superelastic properties even if the archwire is not superelastic. Another limitation of this type of testing is that it does not truly reflect an archwire ligated in a bracket slot and the friction that would have to be overcome with the use of ligatures.

When carrying out bending tests with superelastic orthodontic archwires it is important that the archwire is deflected more than 2 mm. If the archwire is deflected less than this, stress-induced martensitic transformation does not occur meaning the archwire remains mostly in the stiffer austenitic form. This

can affect the results of the studies and the comparability of the properties of the archwires (Nakano *et al.*, 1999; Oltjen *et al.*, 1997; Segner & Ibe, 1995).

### **1.6.2 Tensile testing**

Tensile testing is generally carried out by attaching the archwire at one end and stretching it. This type of testing was deemed to be the most acceptable form of testing in the early experiments looking at the mechanical properties of archwires. Tensile forces are not commonly seen in orthodontic practice. A stretch greater than 8% is required to elicit the expression of superelastic properties when examining superelastic archwires (Asgharnia & Brantley, 1986; Burstone *et al.*, 1985; Miura *et al.*, 1986). Tensile tests tend to produce lower modulus of elasticity values than testing carried out through bending (Asgharnia & Brantley, 1986).

### **1.6.3 Variation in nickel titanium archwires**

There are many brands of archwires currently available to orthodontists. The mechanical properties of the archwires are often not available thus making comparison between manufacturers difficult. Additionally, very often the testing methods and conditions differ which again make direct comparisons difficult. Differences have also been noted in the literature when comparing archwires from different batches from the same manufacturer. Various studies have been carried out in recent times looking at inter- and intra-brand differences in mechanical properties. Bellini *et al.* (2016) tested the

mechanical properties of five 0.016" nickel titanium archwires taken from each of two batches from eight manufacturers (a total of 80 archwires), using standardised testing methods (3-point wire bending test at 36 °C), as outlined in ISO 15841:2006. The 10 mm straight wire spans were centrally loaded to a distance of 3.1 mm and were released at the same speed. In terms of determining inter-brand differences, force was compared at 0.5-, 1-, 2- and 3- mm deflections as well as comparing the slope of the unloading plateau. The maximum force imparted by the archwires at a displacement of 3 mm ranged from 1.7N to 2.3N and there was a statistically significant difference between the brands for all displacement measurements, except for 3 mm displacements. Inter-batch differences were also evaluated using the two batches included for each brand and statistically significant differences were noted between the batches of three brands. The sample size of archwires tested fell below the six archwires per batch recommended in ISO 15841:2006. However, this was a well-controlled laboratory-based study.

Another study which followed the ISO 15841:2006 standard is that carried out by Brauchli *et al.* (2011). In this study three-point bending tests, carried out at 36 °C, were completed on three 0.016" x 0.022" archwires (Titanol, Neo Sentalloy and Copper Nickel Titanium). The straight portion of the distal end of the archwire, measuring 10 mm, was used for testing. The samples were loaded to a maximum deflection of 3.1 mm using an Instron 3344 machine. Force- deflection diagrams were created and force levels at the midpoint of the unloading curve (1.5 mm) were chosen for the comparison of archwires. This study compares how heat treatments affect the archwire

which is not directly relatable to this study but the method of comparing the unloading forces between archwires is useful.

Nakano *et al.* (1999) studied 42 brands of nickel titanium archwires from 9 manufacturers subdivided into 19 brands of 0.016" archwires and 23 brands of 0.016" x 0.022" archwires. Three-point wire bending tests in a controlled environment was carried out to investigate the properties of the archwires from the different brands. The supports for the archwire were 14 mm apart and the length of the wire being tested was not specified. Each sample was tested at 37 °C and was centrally loaded to a displacement of 2 mm. This was repeated three times for each brand using a new wire each time. The amount of force varied greatly between brands tested with 3 point bending tests. Of note, there was a difference of 136g observed between brands in the maximum force delivered when the 0.016" archwires were displaced by 2 mm. A combination of superelastic and conventional nickel titanium archwires were tested in this study which may have influenced the results. Archwires were not displaced beyond 2 mm meaning that it is possible that the superelastic behaviours of the archwire were not activated through stress-induced martensitic transformation.

## **1.7 3D-Printed Models**

### **1.7.1 Manufacturing methods**

Three-dimensional printed models are becoming more popular in orthodontic practice as digital workflows are increasingly adopted and refined. Subtractive or additive manufacturing processes can be used depending on the technology

employed. Additive processes involve a layering procedure whereby additional material is laid down while the subtractive process involves a milling type procedure from a prefabricated block of material. The additive process is considered to be more sustainable and efficient due to the high volume of wastage associated with the subtractive process. Additionally there are a number of different manufacturing methods of 3D printing. The most common being a stereolithography apparatus (SLA) or masked stereolithography apparatus (MSLA). Instead of using a laser beam to trace the layers as used in the SLA, the MSLA process uses a photopolymer resin which is illuminated by an ultraviolet light source through a screen that displays images of the layers, known as masks. A vat with resin is installed and the platform is submerged into the resin to the depth of one layer. Once the layer is illuminated and cured the platform rises and then lowers down clearing space for another layer. The cycle continues until the model is complete in layers. The light wavelength used in this machine is 405 nanometers which has become a standard for MSLA printers. The resolution for MSLA is dependent on the resolution of the LCD screen with the Sonic XL 4K print volume being 120 x 190 x 200 millimeters. The resolution is the same across the whole area meaning that multiple models can be printed simultaneously. The curing light module uses LCD 3.0 technology and comprises a matrix of ultraviolet LEDs with special lenses over them providing uniform curing over the entire print area. Once printed, the supports needed to prevent deflection due to gravity, are removed from the models. They are then washed in isopropyl alcohol and are placed under UV light for further curing as approximately only 80% of the total

polymerization will occur within the 3D printer. (Dawood *et al.*, 2015; Turkyilmaz & Wilkins, 2021).

## **1.8 Previous studies evaluating archwire performance in typodont simulations**

### **1.8.1 Permanent deformation of archwires: Typodont simulations**

Andresen & Barrett (1973) used a typodont set up to compare the performance of 55 cobalt substituted nitinol archwires with stainless steel and Twist-Flex™ archwires in the initial aligning phase of treatment. A selection of archwire dimensions were included giving 8 subgroups containing 10 archwires to be tested: annealed and unannealed cobalt substituted nitinol (0.019"), stainless steel (0.014", 0.016", 0.018") and Twist-Flex™ (0.0175", 0.0195", 0.0215"). The percentage return of the archwire to the original form after being tied into malaligned typodont teeth for 1 hour was evaluated. The typodont was set up so that every second tooth was displaced more than the previous, giving 6 points of displacement (1.4-, 1.9-, 1.9-, 2.8-, 3.3-, 3.7- mm). Each archwire was tied into Siamese Standard edgewise brackets with a steel ligature using a tying pliers. After 1 hour, the archwire was released and a photo was taken from a standardised distance. A Helios vernier callipers with a dial read-out of 1/20 mm was used to measure the distance of the released wire to the deepest point of the bracket and was compared to the distance of an unused wire to the deepest point of the bracket. Two examiners completed these measurements and they were found to have good inter-examiner agreement. A statistically significant difference in the performance of cobalt

substituted nitinol when compared with stainless steel and Twist-Flex™ in returning to the original archform was identified when tooth displacements were greater than 1.5 mm. Multiple deformations were measured, which may affect the accuracy in determining the deformation at given points due to the cumulative error of multiple deformations. The testing conditions were not specified. The archwires were ligated in the brackets for one hour, which reflects deformation at initial ligation. However, it is possible that the degree of deformation could increase as a function of time, which was found in later studies (Hudgins *et al.*, 1990).

Similar limitations were noted by Barrows *et al.* (1982) comparing 5 archwires with similar stiffness to see which has the greatest working flexibility range. Six preformed ideal archforms of each of the 5 different types of archwire were ligated into the brackets of typodont teeth. Every other tooth in the typodont was placed in the ideal archform while the alternative teeth were displaced lingually at increasing depths of 1 mm. An ultra-fine permanent marker was used to mark measuring points on the archwire. The archwire was left in place for 1 hour before being released and the distortions were measured with a calliper calibrated to 0.01 inches.

Hudgins *et al.* (1990) attempted to quantify permanent deformation of nickel titanium after short- and long-term deflection in a laboratory-based study using nine 0.016" archwires from different manufacturers. A stainless steel and a titanium molybdenum alloy archwire were also included to allow comparison. The archwires were deflected into self-ligating brackets of a simulated archform with the maxillary lateral incisor positioned 5 mm palatally and an inter-bracket span of 19 mm between the canine and central incisor.

Deformation of the archwire was measured following 1, 14 and 28 days of ligation, while stored in a water bath at 37 °C, by measuring the distance of the lingual edge of the wire to the facial surface of the bracket before being engaged into the bracket slot and subtracting the repeated measurement after the wire was released from the bracket of the lateral incisor. A new wire was used for each time interval. Deformation was measured using a microscope to  $\pm 0.04$  mm. It was noted that after the archwire was initially released, more springback occurred afterwards in the range of 0.04 to 0.08 mm. This is known as the delayed rebound effect and to allow for this, measurements were taken 1 hour after being released. The degree of deformation increased over time, from 1 day to 14 days to 28 days, although not to a statistically significant effect. A higher degree of deformation was noted with nitinol, stainless steel and titanium molybdenum alloy archwires than with the newer nickel titanium archwires as would be expected. Another interesting result from this study was that there were similar deformation measurements after 1 day and 28 days of ligation. Therefore, a recommendation from this study was that further studies should deploy 1-day time intervals rather than ligating the wire for extended amounts of time. Hudgins *et al.* (1990) also noted that only wire deformation of approximately 1 mm would be considered to be clinically significant.

### **1.8.2 Permanent deformation of archwires: 3-point bending tests using typodonts**

Gurgel *et al.* (2000) considered the force deflection of eight 0.017" x 0.025" nickel titanium archwires in a controlled laboratory setting using a plastic typodont representing mandibular teeth with a moveable displaceable lateral incisor tooth. An inter-bracket distance of 5 mm was chosen and 0.018" x 0.025" brackets were applied. Testing was completed within a chamber at 35 °C. The archwire sample consisted of copper nickel titanium, nitinol, superelastic nickel titanium, heat-activated nickel titanium and Neo Sentalloy archwires. Ligation was with elastomeric modules. Five archwires from each batch were deflected from 0.2 mm to 2 mm using an Instron testing machine. The mean forces of activation and deactivation, recorded at 1 mm displacement, were selected for comparison and were represented on load deflection graphs. All of the archwires tested showed superelastic behaviour but there were differences in the force levels throughout the unloading plateaus. Nickel titanium archwires with different compositions were used which is likely a contributing factor to this variation.

Wilkinson *et al.* (2002) used a modified bending test to investigate the load deflection characteristics of seven 0.016" aligning archwires. The sample included one multistranded stainless steel archwire, one superelastic nickel titanium archwire and five brands of heat-activated nickel titanium archwires. Five testing models were created and used to simulate a displaced lateral incisor tooth. The inter-bracket distance used for all models in this study was derived from typical tooth dimension sizes for a male dentition

(Moyers, 1976). Model 1 consisted of a brass jig with supports 15.5mm apart, representing the inter-bracket distance between a central incisor and canine tooth and was used for 3-point bending tests. Model 2 consisted of a modified acrylic block with Mini-Diamond brackets (conventional brackets using elastomeric modules) while model 3 was the same but with Twin-Lock brackets (self-ligating brackets). Both had four brackets attached representing teeth 11, 21, 23 and 24 and space was left to allow deflection of the archwire at the left lateral incisor site. Model 4 and 5 consisted of a solid acrylic archform onto which Mini-Diamond and Twin-Lock brackets were bonded respectively. An Instron 1193 universal testing machine was used to deflect the wire in buccolingual direction for the flat acrylic models and occluso-gingivally for the typodont model. The load deflection characteristics of the archwires were investigated at temperatures of 22 °C, 35.5 °C and 44 °C to deflections distances of 1-, 2 -, 3 - and 4- mm. A ranking system was used to compare the archwires rather than interpreting the actual data due to questions over the validity in inferring numerical load values to clinical situations.

Similar to the study by Wilkinson *et al.* (2002), Lombardo *et al.* (2012) looked at the load deflection characteristics of superelastic and thermally-active nickel titanium archwires. They looked specifically at the unloading plateau for each archwire and examined the average plateau force, length and slope. They tested 48 archwires with diameters of 0.010" to 0.016" from seven manufacturers. The straightest distal end of the archwire was used for testing. In this study, four self-ligating brackets were attached to a resin base with a 14 mm span between the middle brackets. Archwires were tested at constant

temperature in a water bath at 37 °C. A metal blade attached to the load cell of an Instron 4467 machine was used to deflect the archwire to a distance of 4 mm. They noted an increase in the average plateau unloading force as the archwire increased in diameter from 0.012" to 0.014" or 0.016" and from 0.014" to 0.016".

### **1.8.3 Studies evaluating the force levels imparted by orthodontic archwires**

Higa *et al.* (2017), considered the deflection force of orthodontic archwires ligated in self-ligating brackets and conventional brackets with elastomeric or metal ligatures. Conventional and thermally-activated 0.014" and 0.016" nickel titanium archwires were tested, using 10 archwires per sample group. The sample of 10 archwires were used to minimise the chance of technical errors occurring. A clinical simulation device was created using an acrylic resin sheet with a parabolic shaped trough. This allowed placement of fixed metal rods with brackets attached to acrylic blocks at 6mm intervals, representing the teeth. The exception was the rod representing the upper right central incisor which was free to move in a buccolingual direction. Archwires were ligated to the brackets and the device was suspended in a water bath at 36.5 °C. An Instron 3342 Universal Testing Machine with a load cell of 10N was used to determine the force levels. The archwire was deflected to 3.1 mm representing a displaced tooth at the beginning of orthodontic treatment and the unloading force was measured at 3-, 2-, 1- and 0.5- mm intervals. The mean force imparted increased as the wire was deflected up to 3 mm. In

comparing 0.014" and 0.016" archwires, a higher force was imparted by the 0.016" archwires except for the conventional brackets with metal ligation at 0.5 mm deflection. When archwires with small dimension are being tested, self-ligating bracket systems tend to be associated with lower forces compared to conventional bracket system. The reason for this is that in conventional bracket use, the ligation mode pushes the wire into the bracket slot which increases friction as well as increasing the amount of deflection of the archwire whereas in self ligating brackets the wire is not pressed inside the slot walls.

In separate findings, the load deflection characteristics of an archwire was deemed to be unaffected by the direction of loading either buccolingually or occluso-gingivally (Rock & Wilson, 1988; Schaus & Nikolai, 1986).

## **1.9 Temperature control**

Temperature control is important when testing orthodontic archwires. Equipment such as insulated chambers with infrared lighting, water baths at a constant temperature or thermostats have been used in previous studies (Santoro & Beshers, 2000; Serrano *et al.*, 2014). Serrano *et al.* (2014) used a temperature-controlled chamber with infrared light lamp at  $37 \pm 2$  °C to simulate the oral cavity. This was the same temperature used by Wilkinson *et al.* (2002). However, Moore *et al.* (1999) felt that 35.5 °C was a more appropriate temperature for orthodontic archwires *in vitro*. It is possible to carry out an experiment at room temperature except when testing thermoelastic nickel titanium archwires with a temperature transitional range

of 35 °C. This would affect the results in terms of the force imparted. At room temperature, the archwires would be more flexible than they would be at intra oral temperature as it is below the transition temperature of 35 °C meaning the force levels recorded will be lower and would not reflect intra oral force levels (Kusy, 1997).

### **1.10 Literature Review Summary**

Based on the current available literature, studies looking at the properties and deformation of orthodontic archwires have mostly used 3-point bending tests. These laboratory tests unfortunately do not closely resemble clinical scenarios. To overcome this issue, certain studies have tested the archwires using a variety of clinical simulations including brackets attached to acrylic blocks or typodont set ups with brackets attached to acrylic teeth or rods.

Some studies have attempted to quantify archwire deformation. However, very often they have used typodonts with multiple displaced teeth. While this may resemble a more common clinical scenario of dental crowding, taking multiple measurements from the same archwire may lead to compounding measurement errors leading to inaccuracies in the reporting of deformation. Additionally, studies that examined the extent of archwire deformation have, in the majority of cases, reported on the degree of deformation only. No attempt has been made to group these archwires into deformation categories to help clinicians decide whether an archwire is clinically usable or not.

In terms of archwire selection, most studies included in this review have compared the properties and deformation of archwires composed of different

materials as well as archwires of varying dimensions. The variation in these studies make it difficult to compare the archwire deformation between one study and another. It was noted also that variation exists in terms of archwire properties and degree of deformation between archwires from the same manufacturer as well as those from different manufacturers.

Therefore, a laboratory based study that closely resembles a clinical scenario while reducing the amount of variables within the study is justified. The creation of a validated index is important to categorise archwires based on their degree of deformation after their engagement into brackets attached to displaced teeth.

## **Chapter 2 – Aims, Objectives and Null Hypotheses**

### **2.1 Aims**

- To compare permanent deformation of superelastic nickel titanium aligning archwires produced both by the same and different manufacturer when engaged in teeth with varying displacement in a simulated set up
- To compare the maximum force imparted by these archwires at a variety of magnitudes of tooth displacement

### **2.2 Objectives**

- To objectively measure the degree of deformation of the archwires by superimposing the used wire onto an image of the original wire before engagement.
- To develop a validated index in order to categorise the used archwires according to the extent of deformation with regard to their suitability for clinical use.
- To objectively measure the force imparted on a tooth at varying increments of displacement.

### **2.3 Null Hypotheses**

- There is no difference in the archform of superelastic nickel titanium archwires before or after they are engaged in the brackets on displaced

lateral incisor teeth of varying increments of displacement (in terms of objectively measured deformation or suitability for clinical use).

- There is no difference in the nature or degree of distortion elicited by a range of superelastic nickel titanium archwires following engagement in a fixed appliance bonded to displaced teeth.

## **Chapter 3 - Materials & methods**

### **3.1 Archwire materials**

Superelastic nickel titanium archwires, with diameters of 0.012" and 0.014", from two different manufacturers were selected. These are commonly used as initial aligning archwires especially where large contact point discrepancies exist. Specifically, the superelastic archwires used were:

1. Euroform® Plus Super Elastic Nickel Titanium Archwires (0.012").

Batch numbers:

- i. 1723 1213
- ii. 1726 1210
- iii. 1722 1214

2. Euroform® Plus Super Elastic Nickel Titanium Archwires (0.014").

Batch numbers:

- i. 1711 1416
- ii. 1720 1414
- iii. 1722 1412

3. American Orthodontics Super Elastic Nickel Titanium Archwires (0.012").

Batch numbers:

- i. P0065305
- ii. P0124422
- iii. P0127966

4. American Orthodontics Super Elastic Nickel Titanium Archwires (0.014").

Batch numbers:

- i. P0046588
- ii. P0065307
- iii. Q15195

### 3.2 Creation of 3D models

3-D printed digital models were created using a Phrozen Sonic XL machine, powered by MSLA or LCD SLA technology, with 4K XY resolution and 0.05 mm layer height (Figure 3.1). The material used in creating the models was FotoDent® model2 385 nm (Dreve, Germany). The models were set up digitally with all teeth aligned except the upper left lateral incisor which was displaced at 3-, 4- or 5- mm increments. Conventional Gemini™ (3M Unitek) brackets with MBT prescription 0.022" x 0.028" slot size were applied to the teeth digitally. Eight identical models were printed for each increment.



**Figure 3.1. 3D-printed models with palatal displacement of the upper left lateral incisor by 3 mm, 4 mm and 5 mm**

### **3.3 Part I - Determination of archwire deformation**

#### **3.3.1 Sample size**

The sample size was not based on any assumptions or statistical estimations. Given the nature of the study design used, with each archwire being ligated for 24 hours, the feasibility and costs were balanced with requirements for statistical power. The ISO 15841 standard for testing orthodontic archwires recommends testing 6 wires from each batch. With this in mind, 8 archwires from each batch of 10 were selected at random for evaluation. Therefore, with 36 subgroups from two manufacturers, two wire gauges and three increments of displacement, a total of 288 archwires were required overall. When examining the load deflection characteristics of the archwires, the remaining two archwires from each batch of 10 archwires were selected for testing requiring a further 72 archwires.

#### **3.3.2 Measurement of uniformity within batches**

In total, 144 each of 0.012" and 0.014" dimensions were evaluated. Eight archwires from three batches from each of the two manufacturers at 3- , 4- and 5- mm increments were tested totaling 36 batches (Figure 3.2).



**Figure 3.2. Summary of archwire selection. Nickel titanium archwires with diameters of 0.012" and 0.014" were selected. Eight archwires were tested from three batches of two brands (Euroform or American Orthodontics) for displacements of 3-, 4- or 5- mm increments.**

The examiner was given a numbered envelope containing a batch of archwires and was therefore blinded to the manufacturer of the archwires being tested. Each archwire was assigned a number and a photocopy was taken of the batch of archwires (Figure 3.3). Archwire number 10, selected at random, was considered the control within each batch. Each archwire from the particular batch was placed on top of the photocopy of the control archwire and conformity with the control was confirmed at 5-fold magnification. This ensured uniformity within the batch before starting the experiment.

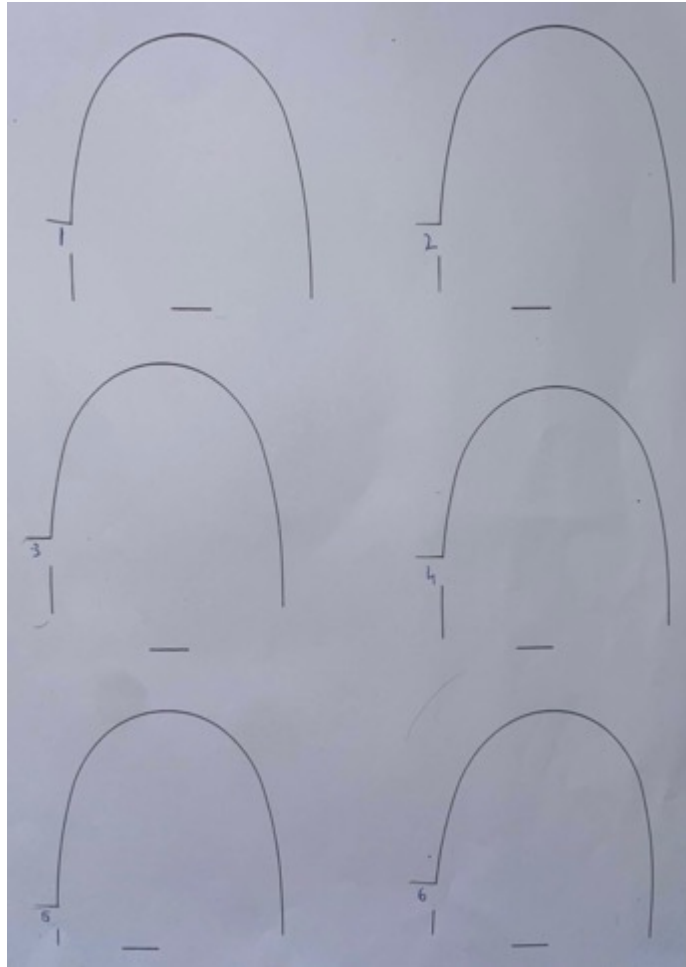


**Figure 3.3. Each archwire was assigned a label (1-10), a photocopy was taken of each batch of archwires and a 1 cm line was drawn to allow calibration of the measuring software, Image J™**

### 3.3.3 Archwire ligation

The archwire was ligated in place with elastomeric modules (Ortho-Care, West Yorkshire, UK) in a standard configuration on each tooth except for the upper left lateral incisor where a figure-of-8 configuration was used. Finger pressure only was used to hold the archwire in the bracket slot during ligation to prevent any inadvertent deformation that might occur if using dental instruments.

### 3.3.4 Degree of deflection of the archwire into the bracket

Once ligated, a photograph was taken using a camera (Canon 250D) on a stand at a set distance of 35 cm. The photo was uploaded to Image J™ software. The angle of the archwire exiting the brackets of adjacent teeth as well as the angle the wire makes as it is deflected into the bracket mesially and distally were measured (Figure 3.4).

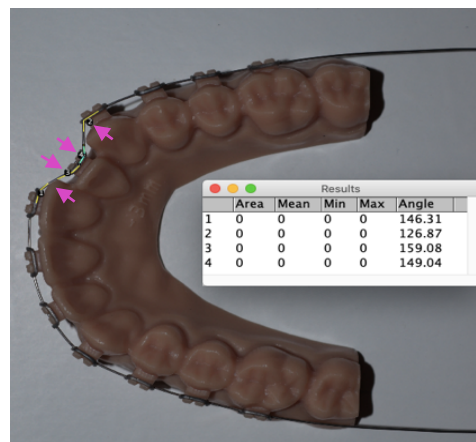


**Figure 3.4. Angles of deflection of the archwire into the displaced tooth measured using Image J™ Software**

### **3.3.5 Testing conditions**

The ISO 15841 standard stipulates the conditions for testing of the mechanical properties orthodontic archwires. Superelastic nickel titanium archwires are tested at  $36 \pm 1$  °C. In line with previous research (Hudgins *et al.*, 1990), the archwires were left in place for 24 hours while being stored in an incubator at  $36 \pm 1$  °C. The temperature was confirmed using two digital thermometers. After 24 hours, the archwires were removed from the models. The archwires were then placed back in the incubator for 1 hour at  $36 \pm 1$  °C to allow for the delayed rebound effect.

### **3.3.6 Measuring the degree of deformation**

#### **3.3.6.1 Subjective measurement of deformation**

An index was developed to categorise the archwires according to their visual appearance once released from the typodont. Three categories were chosen as a higher number of categories would result in less credible results and would be more difficult to apply to a clinical setting. The qualitative categories serve to indicate whether a used archwire conforms to its original archform or is permanently deformed, and, if so, whether or not it is clinically usable.

Each archwire was categorised as follows:

#### **1. No deformation:**

No visible deformation of the wire resembling an unused archwire following the original archform

## 2. Minor deformation:

Minimal irregularities defined by having small kinks while still having a smooth continuous archform. It has not returned completely to its original form but is deemed to remain clinically useable.

## 3. Major deformation:

Definite irregularities defined by visible kinks. There may be gross deformation of the arch form when compared to a template of the original archwire. Due to the extent of the irregularity, it is deemed clinically unusable.

### **3.3.6.1.2 Validation of index**

An operator, not directly involved in this study, created deformations of varying degrees in twenty 0.014" nickel titanium archwires. These archwires were used as part of a calibration exercise to validate the index. The lead examiner as well as two Consultant-level orthodontists were given the index and were asked to categorise the archwires as 1, 2 or 3 depending on the perceived level of deformation. The lead examiner repeated the categorisation of the same archwires after 2 weeks. A weighted kappa co-efficient was used to assess inter- and intra-rater agreement.

### **3.3.6.2 Objective measurement of deformation**

On removing the archwires from the incubator, each archwire was categorised in terms of deformation. The archwire was placed on the photocopy of the

original unused archwire. A photograph was taken using a camera (Canon 250D) on a stand at a set distance of 35 cm above the wire. The image was uploaded to Image J™ software and the image was calibrated using a 1 cm ruler mark. The angle of deformation was measured (Figure 3.5(a)). The length of deformation from the most distal point on the used archwire to the original archwire was measured (Figure 3.5(b)). All measurements were made by a single examiner.

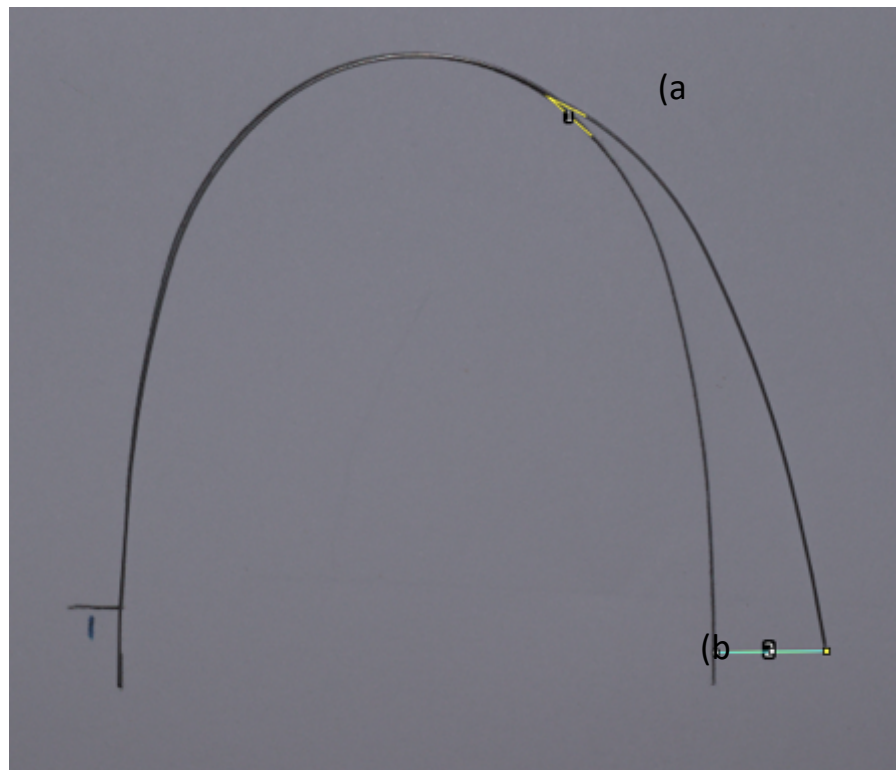


**Figure 3.5. Objective measurements of deformation (a) angular measurement (b) linear measurement**

#### **3.3.6.2.1 Intra-examiner reliability**

The lead examiner re-measured the angular and linear deformation of 56 archwires five times to determine intra-examiner reliability.

#### **3.3.6.3 Statistical analysis**

A weighted kappa coefficient of agreement was used in determining inter- and intra- rater agreement. Using the Shapiro-Wilk normality test, the majority of the data collected for Part I was not normally distributed. On this basis, non-parametric statistical testing was used. Kruskal-Wallis Testing was used for multiple comparisons of wire response in terms of deformation angle and deformation length for each increment of displacement (3 mm vs 4 mm vs 5 mm) for each of the wire dimensions (0.012" and 0.014") as well as in terms of the brands (Euroform and American Orthodontics). Mann-Whitney U Tests were used to compare the archwire dimensions (0.012" vs 0.014") and archwire brands (Euroform vs American Orthodontics).

### **3.4 Part II - Determination of force delivered by archwires**

#### **3.4.1 Wire bending tests**

Modified 3-point wire bending tests were carried out on a total of 72 nickel titanium archwires of 0.012" and 0.014" dimensions. The 3D-printed models were used to replicate clinical scenarios rather than using a straight portion of an archwire. These tests were carried out using a Universal Testing Machine (Zwick Roell, ZwickiLine Z5.0) with 0.1N resolution (Figure 3.6). A silicone jig

was constructed to secure the archwire within the load cell. A clamp was used to hold each testing model in the same position within the machine (Figure 3.7). Each archwire being tested was ligated, using elastomeric modules (Ortho-Care, West Yorkshire, UK), in all brackets except the upper left lateral incisor.



**Figure 3.6. Zwick Roell ZwickiLine Z5.0 used for three-point bending test**

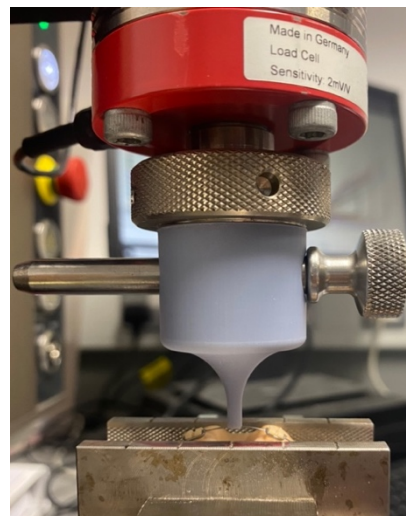


**Figure 3.7. Testing model with archwire ligated secured within the clamp and blue silicone jig to accommodate the archwire during testing**

Two unused archwires (archwire number 9 and 10) from each batch were selected for testing. Each archwire was deflected at three increments representing displacement of the upper left lateral incisor by 3-, 4- or 5- mm. The displacement limit was set within the machine. Each archwire went through three cycles of loading and unloading at each increment. Testing was conducted at  $22 \pm 1$  °C. The ligated archwires were loaded at the midpoint of the inter-bracket space between the upper left central incisor and canine. The wire was deflected at a crosshead speed of 25 mm/min until deflection of 3-, 4- or 5- mm was achieved. Each archwire was unloaded at the same crosshead speed until the force returned to zero.

### **3.5.2 Data collection**

Loading and unloading forces were recorded directly using a proprietary software programme. This data was exported to Microsoft Excel. From this, a graph for visual comparison was plotted for each test showing deflection of the wire, reported in millimetres, on the x-axis and the force, measured in Newtons, on the y-axis.

### **3.5.3 Statistical analysis**

Independent samples Mann-Whitney U Tests were used to compare the maximum force exerted between brands and archwire sizes. The Kruskal Wallis test for non-parametric data was used to evaluate between the batches within brands as there were more than two batches within each brand.

## **Chapter 4 – Results**

### **4.1 Part I - Determination of archwire deformation**

#### **4.1.1 Measurement of uniformity within batches**

In total, 36 batches of archwires were tested. Within each batch, when compared to the control, the archform of every archwire was identical to the control. This confirmed uniformity with the batches prior to use.

#### **4.1.2 Degree of deflection of the archwire into the bracket**

Table 1 summarises the results for the angulation of the archwire leaving the brackets of the adjacent teeth (upper left central incisors and upper left canine) entering the upper left lateral incisor bracket. As expected, the angle of deflection became more acute as the degree of tooth displacement increased.

**Table 4.1. Deflection angles of the archwire engaged at displacement increments of 3-, 4-and 5- mm**

|                                                 | <b>3 mm</b> | <b>4 mm</b> | <b>5 mm</b> |
|-------------------------------------------------|-------------|-------------|-------------|
| <b>Archwire leaving bracket mesially</b>        | 146°        | 139°        | 129°        |
| <b>Archwire leaving bracket distally</b>        | 127°        | 126°        | 125°        |
| <b>Deflection of wire into bracket mesially</b> | 159°        | 153°        | 132°        |
| <b>Deflection of wire into bracket distally</b> | 149°        | 143°        | 126°        |

### **4.1.3 Subjective measurement of deformation**

#### **4.1.3.1 Calibration of the examiner**

A weighted kappa coefficient score of 0.698 was achieved for inter-rater agreement in the overall categorisation of the archwires at 1, 2 or 3 depending on the degree of wire deformation, demonstrating a substantial level of agreement. The level of agreement varied between categories. The inter-rater agreement score was 0.712 for archwires categorised as 1, 0.574 for those categorised as 2 and 0.792 for those scoring 3. A score of 1 was achieved for intra-rater agreement, showing an almost perfect level of agreement (Landis & Koch, 1977).

#### 4.1.3.2 Categorisation of the archwires

Tables 4.2 and 4.3 show the categorisation results for 3-, 4- and 5- mm displacements across both brands (Euroform and American Orthodontics) and archwire dimensions (0.012" and 0.014").

**Table 4.2. Deformation of 0.012" Euroform and American Orthodontics archwires at 3-, 4-, 5- mm displacements**

| Wire Details   |              |       | Deformation Category (1, 2, 3) |       |       |        |
|----------------|--------------|-------|--------------------------------|-------|-------|--------|
| Wire Thickness | Displacement | Brand | 1 (%)                          | 2 (%) | 3 (%) | Median |
| 0.012"         | 3 mm         | EF    | 58%                            | 33%   | 8%    | 1      |
| 0.012"         | 3 mm         | AO    | 42%                            | 58%   | 0%    | 2      |
| 0.012"         | 4 mm         | EF    | 58%                            | 42%   | 0%    | 1      |
| 0.012"         | 4 mm         | AO    | 46%                            | 33%   | 21%   | 2      |
| 0.012"         | 5 mm         | EF    | 13%                            | 54%   | 33%   | 2      |
| 0.012"         | 5 mm         | AO    | 29%                            | 63%   | 8%    | 2      |

**Table 4.3. Deformation of 0.014” Euroform and American Orthodontics archwires at 3-, 4-, 5- mm displacements**

| Wire Details   |              |       | Deformation Category (1, 2, 3) |       |       |        |
|----------------|--------------|-------|--------------------------------|-------|-------|--------|
| Wire Thickness | Displacement | Brand | 1 (%)                          | 2 (%) | 3 (%) | Median |
| 0.014”         | 3 mm         | EF    | 63%                            | 33%   | 4%    | 1      |
| 0.014”         | 3 mm         | AO    | 50%                            | 38%   | 13%   | 2      |
| 0.014”         | 4 mm         | EF    | 67%                            | 21%   | 13%   | 1      |
| 0.014”         | 4 mm         | AO    | 21%                            | 67%   | 13%   | 2      |
| 0.014”         | 5 mm         | EF    | 42%                            | 54%   | 4%    | 2      |
| 0.014”         | 5 mm         | AO    | 13%                            | 63%   | 25%   | 2      |

#### **4.1.3.3 Analysis of deformation categories**

##### **4.1.3.3.1 Overall comparison between brands**

The median values for archwire deformation categories were calculated using Microsoft Excel for each brand, archwire thickness and archwire displacement (Table 4.2 & 4.3). 0.012” and 0.014” archwires followed a similar pattern. For archwire displacements of 3 mm or 4 mm, the median archwire score was 1 for Euroform and 2 for American Orthodontics. However, at 5 mm displacement, the median value was 2 for both manufacturers (Tables 4.2 and 4.3; Figure 4.1). Based on the Mann-Whitney U Test, Euroform archwires had lower distortion categories compared to the American Orthodontics archwires ( $P = 0.008$ ).

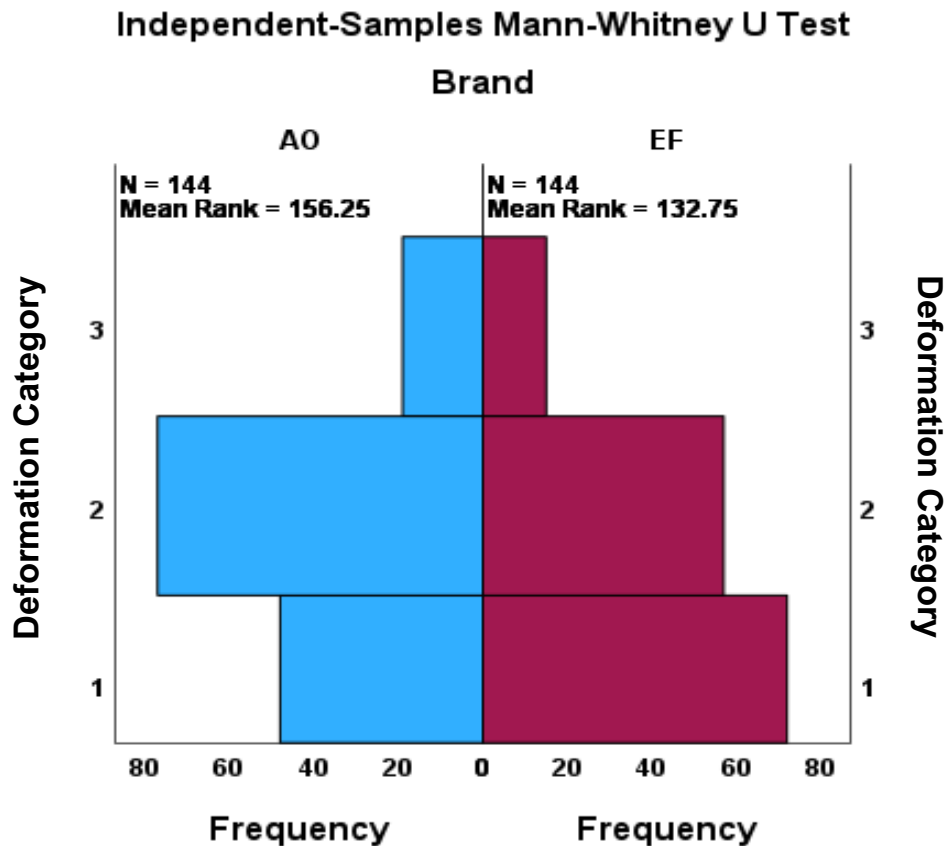


**Figure 4.1. Overall frequency of archwires, including both 0.012” and 0.014” wires, scoring as 1-3 in relation to wire deformation**

#### **4.1.3.3.2 Comparison between archwire brands for different archwire dimensions (0.012” and 0.014”)**

Based on the Mann-Whitney U Test (Figure 4.2), there was no statistically significant difference in distortion noted between 0.012” Euroform and American Orthodontics archwires ( $P = 0.89$ ). However, the 0.014” Euroform

archwire had significantly lower distortion than American Orthodontics archwires ( $P = <0.001$ ; Figure 4.3).

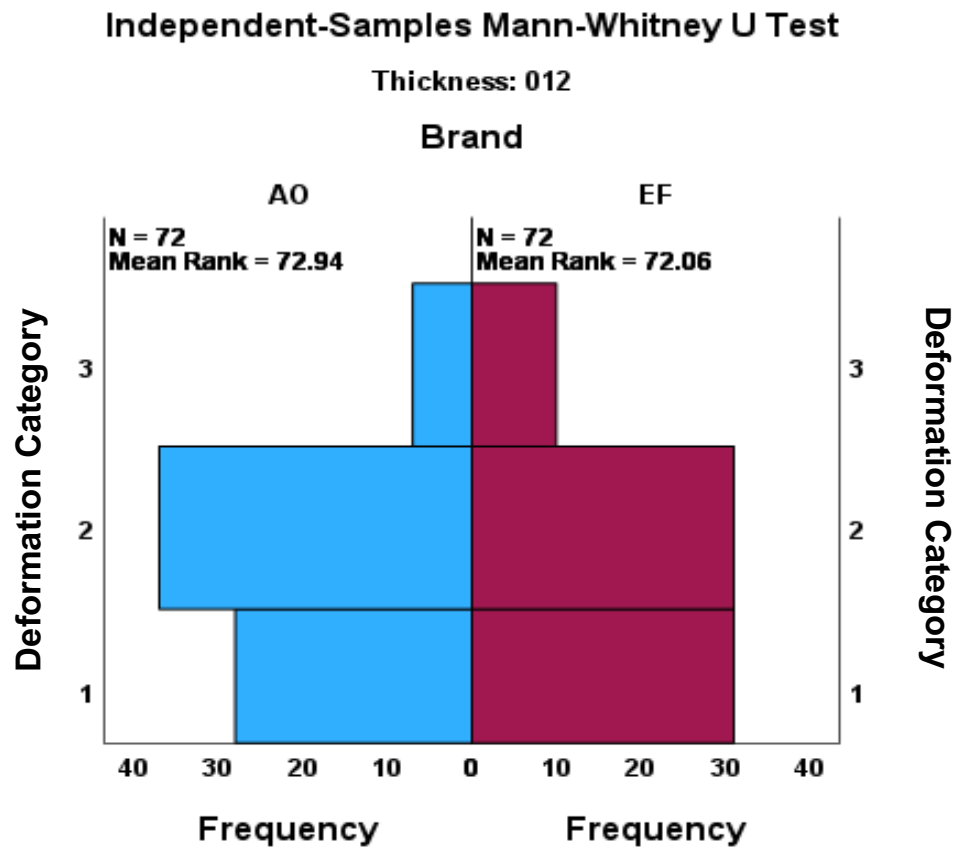


**Figure 4.2. Frequency of 0.012" archwires, including both American Orthodontics and Euroform wires scoring as 1 - 3 in relation to wire deformation**

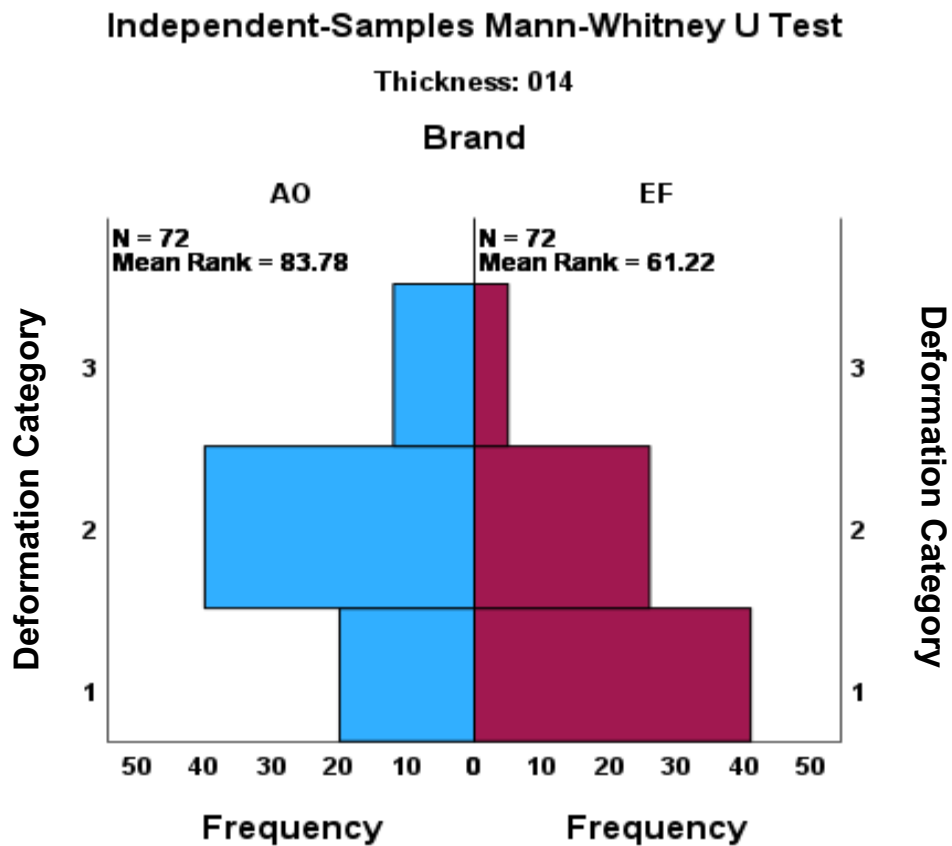


**Figure 4.3. Frequency of 0.014” archwires, including both American Orthodontics and Euroform wires scoring as 1 - 3 in relation to wire deformation**

#### **4.1.3.3.4 Comparison between batches of archwires within each brand**

Independent-Samples Kruskal-Wallis Tests were used to determine if there were any significant differences noted between the batches of wires within each of the brands. There was no statistically significant difference noted

between individual batches for either Euroform ( $P = 0.169$ ) or American Orthodontic brands ( $P = 0.267$ ).

#### **4.1.3.3.5 Intra-batch comparison of archwires**

A statistically significant difference noted between the archwires within two batches - Euroform 1720 1414 and American Orthodontics P0046588 ( $P = 0.035$ ). Both batches contained 0.014" archwires. No statistically significant intra-batch differences were noted in the other 34 batches.

### **4.1.4 Objective measurement of distortion**

#### **4.1.4.1 Intra-examiner reliability**

Overall intraclass coefficient scores of 0.935 and 0.998 were achieved for intra-examiner agreement in measuring the deformation angles and lengths respectively, demonstrating an almost perfect level of agreement. The intra-examiner agreement remained as an almost perfect level of agreement across both the brands and archwire displacements (Table 4.4).

**Table 4.4. Intraclass correlation coefficient calculations for deformation angles and lengths**

|              |      | Deformation Angle | Deformation Length |
|--------------|------|-------------------|--------------------|
| Brand        | AO   | 0.924             | 0.998              |
|              | EF   | 0.952             | 0.995              |
| Displacement | 3 mm | 0.871             | 0.994              |
|              | 4 mm | 0.965             | 0.998              |
|              | 5 mm | 0.924             | 0.998              |
| Overall      |      | 0.935             | 0.998              |

#### 4.1.4.2 Linear measurements

The mean linear results for 0.012" archwires displaced to 3-, 4- or 5- mm from both American Orthodontics and Euroform brands are presented in Table 4.5 and represented by Figure 3.5(a). Pairwise comparison of the mean values showed no statistical difference between 3-, 4- or 5- mm displacements for American Orthodontic archwires (Table 4.6). However, a statistically significant difference was noted for Euroform archwires between 3 mm and 5 mm displacements, with mean deformation lengths of 0.9 mm and 3.1 mm noted respectively ( $P = <0.001$ ), as well as between 4 mm and 5 mm displacements with deformation lengths of 0.8 mm and 3.1mm ( $P = <0.001$ ).

**Table 4.5. Linear measurements of 0.012” archwires (Mean, Standard Deviation [SD], Median, Interquartile Range [IQR]) from Euroform and American Orthodontics brands at 3-, 4- or 5- mm displacements**

| Wire Details<br>(0.012”) |       | Deformation Length (mm) |      |        |             |             |      |
|--------------------------|-------|-------------------------|------|--------|-------------|-------------|------|
| Disp.                    | Brand | Mean                    | SD   | Median | Perc.<br>25 | Perc.<br>75 | IQR  |
| 3 mm                     | AO    | 0.82                    | 0.61 | 0.95   | 0.00        | 1.35        | 1.35 |
|                          | EF    | 0.86                    | 2.07 | 0.00   | 0.00        | 1.10        | 1.10 |
| 4 mm                     | AO    | 2.85                    | 4.38 | 1.00   | 0.30        | 2.45        | 2.15 |
|                          | EF    | 0.76                    | 0.82 | 0.65   | 0.00        | 1.05        | 1.05 |
| 5 mm                     | AO    | 2.22                    | 4.07 | 1.35   | 0.45        | 2.35        | 1.90 |
|                          | EF    | 3.10                    | 2.79 | 2.10   | 1.45        | 4.30        | 2.85 |

The mean magnitude of deformation of 0.014” wires increased when the archwires were displaced to greater distances (from 3 mm to 4 mm to 5 mm) for both manufacturers (Table 4.6). Pairwise comparison (Table 4.7) showed no statistical difference in the mean deformation length in Euroform archwires displaced to 3-, 4- or 5- mm although a statistically significant difference in American Orthodontic wires at 3 mm and 5 mm displacements with mean deformation lengths of 1.8 mm and 4 mm was observed ( $P = 0.002$ ).

**Table 4.6. Linear measurements of 0.014” archwires (Mean, Standard Deviation [SD], Median, Interquartile Range [IQR]) from Euroform and American Orthodontics brands at 3-, 4- or 5- mm displacements**

| Wire Details<br>(0.014”) |       | Deformation Length (mm) |      |        |             |             |      |
|--------------------------|-------|-------------------------|------|--------|-------------|-------------|------|
| Disp.                    | Brand | Mean                    | SD   | Median | Perc.<br>25 | Perc.<br>75 | IQR  |
| 3 mm                     | AO    | 1.82                    | 3.29 | 0.40   | 0.00        | 2.10        | 2.10 |
|                          | EF    | 0.71                    | 1.08 | 0.00   | 0.00        | 1.10        | 1.10 |
| 4 mm                     | AO    | 2.44                    | 3.50 | 1.40   | 0.95        | 2.15        | 1.20 |
|                          | EF    | 1.08                    | 1.66 | 0.55   | 0.00        | 1.20        | 1.20 |
| 5 mm                     | AO    | 4.02                    | 4.32 | 2.50   | 1.60        | 3.45        | 1.85 |
|                          | EF    | 1.35                    | 2.30 | 0.85   | 0.00        | 1.95        | 1.95 |

**Table 4.7. A summary of the statistical analysis of the deformation lengths of 0.012” and 0.014”, including both Euroform and American Orthodontics archwires at 3-, 4- and 5- mm displacements**

| Brand | Diameter | Deformation Length               |             |            |
|-------|----------|----------------------------------|-------------|------------|
| AO    | 0.012"   | Multiple Comparison              |             | 0.167      |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS         |
|       |          |                                  | 3 mm – 5 mm | NS         |
|       |          |                                  | 4 mm – 5 mm | NS         |
|       | 0.014"   | Multiple Comparison              |             | 0.003      |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS         |
|       |          |                                  | 3 mm – 5 mm | (-) 0.002  |
|       |          |                                  | 4 mm – 5 mm | NS         |
| EF    | 0.012"   | Multiple Comparison              |             | < 0.001    |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS         |
|       |          |                                  | 3 mm – 5 mm | (-) <0.001 |
|       |          |                                  | 4 mm – 5 mm | (-) 0.001  |
|       | 0.014"   | Multiple Comparison              |             | 0.459      |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS         |
|       |          |                                  | 3 mm – 5 mm | NS         |
|       |          |                                  | 4 mm – 5 mm | NS         |

#### 4.1.4.3 Angular measurements

The mean angular results for 0.012" archwires displaced to 3-, 4- or 5- mm from both American Orthodontics and Euroform brands are presented in Table 4.8 and represented by Figure 3.5(b). Pairwise comparison of the mean values showed a statistically significant difference in Euroform archwires between 3 mm and 5 mm displacements, with mean deformation angles of 4.4° and 9.1° respectively ( $P = 0.002$ ), as well as between 4 mm and 5 mm displacement with mean deformation angles of 4° and 9.1° ( $P = 0.004$ ).

**Table 4.8. Angular measurements of 0.012" archwires (Mean, Standard Deviation [SD], Median, Interquartile Range [IQR]) from Euroform and American Orthodontics brands at 3-, 4- or 5- mm displacements**

| Wire Details<br>(0.012") |       | Angle of Deformation (degrees) |      |        |             |             |      |
|--------------------------|-------|--------------------------------|------|--------|-------------|-------------|------|
| Disp.                    | Brand | Mean                           | SD   | Median | Perc.<br>25 | Perc.<br>75 | IQR  |
| 3 mm                     | AO    | 5.67                           | 4.01 | 7.00   | 0.00        | 9.00        | 9.00 |
|                          | EF    | 4.33                           | 5.71 | 0.00   | 0.00        | 7.50        | 7.50 |
| 4 mm                     | AO    | 6.75                           | 6.18 | 5.50   | 1.50        | 9.50        | 7.00 |
|                          | EF    | 4.04                           | 3.57 | 3.50   | 0.00        | 7.50        | 7.50 |
| 5 mm                     | AO    | 6.58                           | 4.88 | 7.00   | 5.00        | 8.00        | 3.00 |
|                          | EF    | 9.00                           | 4.70 | 8.50   | 7.00        | 13.00       | 6.00 |

The mean deformation angles of 0.014” archwires increased when the archwires were displaced to greater distances (from 3 mm to 4 mm to 5 mm) for both manufacturers (Table 4.9). However, pairwise comparison showed no statistically significant differences in the deformation angle in Euroform or American Orthodontics archwires (Table 4.10).

**Table 4.9. Angular measurements of 0.014” archwires (Mean, Standard Deviation [SD], Median, Interquartile Range [IQR]) from Euroform and American Orthodontics brands at 3-, 4- or 5- mm displacements**

| Wire Details<br>(0.014”) |       | Angle of Deformation (degrees) |      |        |             |             |      |
|--------------------------|-------|--------------------------------|------|--------|-------------|-------------|------|
| Disp.                    | Brand | Mean                           | SD   | Median | Perc.<br>25 | Perc.<br>75 | IQR  |
| 3 mm                     | AO    | 5.46                           | 5.53 | 6.50   | 0.00        | 8.50        | 8.50 |
|                          | EF    | 3.50                           | 4.09 | 0.00   | 0.00        | 7.00        | 7.00 |
| 4 mm                     | AO    | 6.83                           | 3.41 | 6.00   | 5.00        | 9.00        | 4.00 |
|                          | EF    | 4.00                           | 5.03 | 3.50   | 0.00        | 6.00        | 6.00 |
| 5 mm                     | AO    | 7.67                           | 4.26 | 7.00   | 5.00        | 11.00       | 6.00 |
|                          | EF    | 5.08                           | 5.11 | 6.50   | 0.00        | 8.50        | 8.50 |

**Table 4.10. A summary of the statistical analysis of the deformation angles of 0.012” and 0.014”, including both Euroform and American Orthodontics archwires at 3-, 4- and 5- mm displacements**

| Brand | Diameter | Deformation Angle                |             |           |
|-------|----------|----------------------------------|-------------|-----------|
| AO    | 0.012"   | Multiple Comparison              |             | 0.892     |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS        |
|       |          |                                  | 3 mm – 5 mm | NS        |
|       |          |                                  | 4 mm – 5 mm | NS        |
|       | 0.014"   | Multiple Comparison              |             | 0.395     |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS        |
|       |          |                                  | 3 mm – 5 mm | NS        |
|       |          |                                  | 4 mm – 5 mm | NS        |
| EF    | 0.012"   | Multiple Comparison              |             | < 0.001   |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS        |
|       |          |                                  | 3 mm – 5 mm | (-) 0.002 |
|       |          |                                  | 4 mm – 5 mm | (-) 0.004 |
|       | 0.014"   | Multiple Comparison              |             | 0.448     |
|       |          | Pair-wise Comparison (Direction) | 3 mm – 4 mm | NS        |
|       |          |                                  | 3 mm – 5 mm | NS        |
|       |          |                                  | 4 mm – 5 mm | NS        |

## **4.2 Part II - Force delivered by archwires**

The results of the archwire bending tests are represented in diagrammatic form within the following section and are divided into upper left lateral incisor displacements of 3-, 4- and 5- mm. As noted in paragraph 3.4.1, the tests have been performed on two wires from each batch. It should be noted that, while the quantitative results included in this study reflect all tests performed, in the interest of clarity and qualitative understanding, the charts below only show the results for one wire from each batch. All data was collected in Microsoft Excel and plotted in graph format to facilitate comparison. The force required to deflect the archwire is measured in Newtons (y-axis) while the deflection in millimeters is reflected on the x-axis. The upward slope of the graph for each individual archwire provides information on the rigidity of the archwire – a less vertical slope represents a more flexible archwire. The peak of each graph represents the maximum force exerted by the wire at a given displacement i.e., 3-, 4- or 5- mm. The unloading curve represents the force delivered by the wire to a tooth as the tooth begins to move.

### **4.2.1 Lateral incisor displacement of 3 mm**

The loading gradient varied among the archwires with some showing a steeper gradient while others followed a more characteristic superelastic pattern. There is variation in the force delivered at maximum displacement. A ‘jamming effect’ was noted in one 0.012” American Orthodontics archwire displaced to 3 mm. This resulted in the archwire becoming stuck at the end of the loading phase; associated results for the unloading phase in the following section

should be interpreted with caution. The force imparted by the archwire at maximum displacement and during the unloading phase depending on the cross-sectional diameter, manufacturer, and batch of archwires used are considered.

#### **4.2.1.1 Effect of archwire cross sectional diameter (0.012" vs 0.014") on the maximum force**

The maximum force imparted by 0.012" and 0.014" nickel titanium archwires including both manufacturers (Euroform and American Orthodontics) at 3 mm displacements are presented in Figure 4.4 A & B. There appears to be a wide range in the maximum force imparted for both 0.012" (1.8N – 3.4N) and 0.014" (2N – 4.2N; Table 4.11).

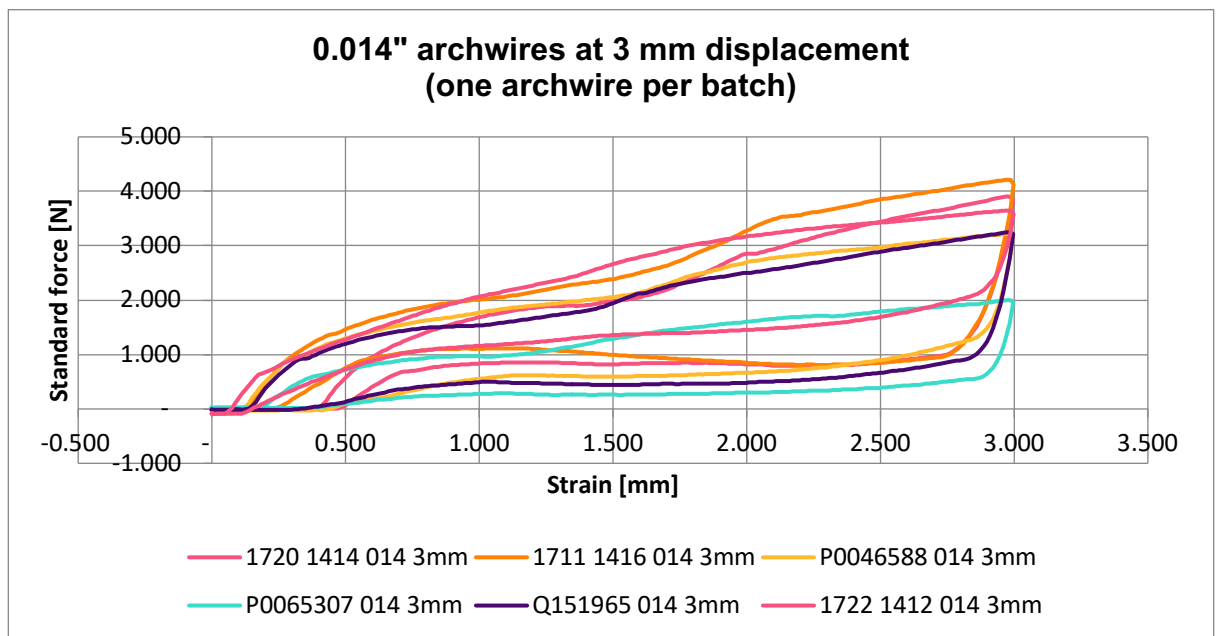
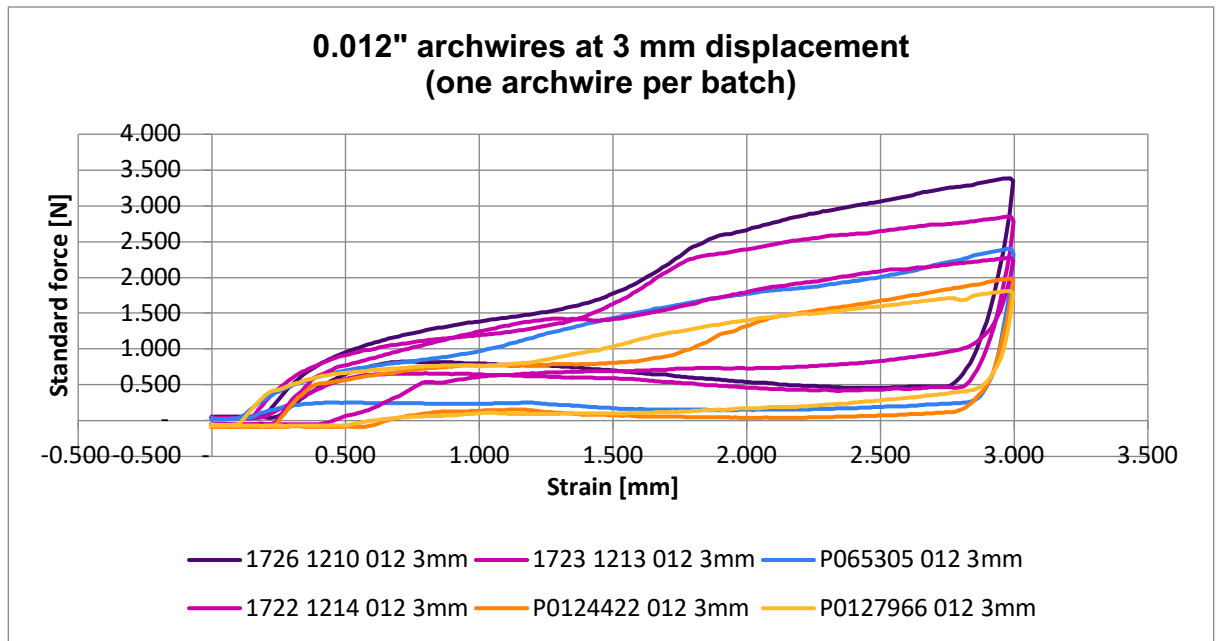


**Figure 4.4 A & B. Load deflection graphs showing the characteristics of 0.012" and 0.014" archwires from both Euroform and American Orthodontics at maximum displacement of 3 mm**

**Table 4.11. The maximum and minimum force (N) for 0.012” and 0.014” archwires (both brands combined) at (a) maximum strain of 3 mm (b) the midpoint, taken at strain of 1.5 mm, of the unloading phase**

|        | Force (N) measured at:      |         |      |                                            |         |      |
|--------|-----------------------------|---------|------|--------------------------------------------|---------|------|
|        | (a) Maximum strain of 3 mm: |         |      | (b) Midpoint of unloading phase at 1.5 mm: |         |      |
|        | Minimum                     | Maximum | Mean | Minimum                                    | Maximum | Mean |
| 0.012” | 1.8                         | 3.4     | 2.4  | 0.0                                        | 0.7     | 0.4  |
| 0.014” | 2.0                         | 4.2     | 3.4  | 0.3                                        | 1.4     | 0.8  |

#### **4.2.1.2 Effect of different archwire manufacturers**

Statistically significant difference in the maximum force imparted at 3 mm displacements between Euroform and American Orthodontics archwires when examined using Independent-Samples Mann-Whitney U Tests ( $P < .001$ ) were observed (Figure 4.5). These differences are demonstrated visually in load deflection diagram format (Figure 4.6 A & B) while the differences in numerical values showing the maximum and minimum forces at maximum strain of 3 mm as well as at the midpoint of the unloading phase are shown in Table 4.12.

# Independent-Samples Mann-Whitney U Test

Max Strain: 3

Brand

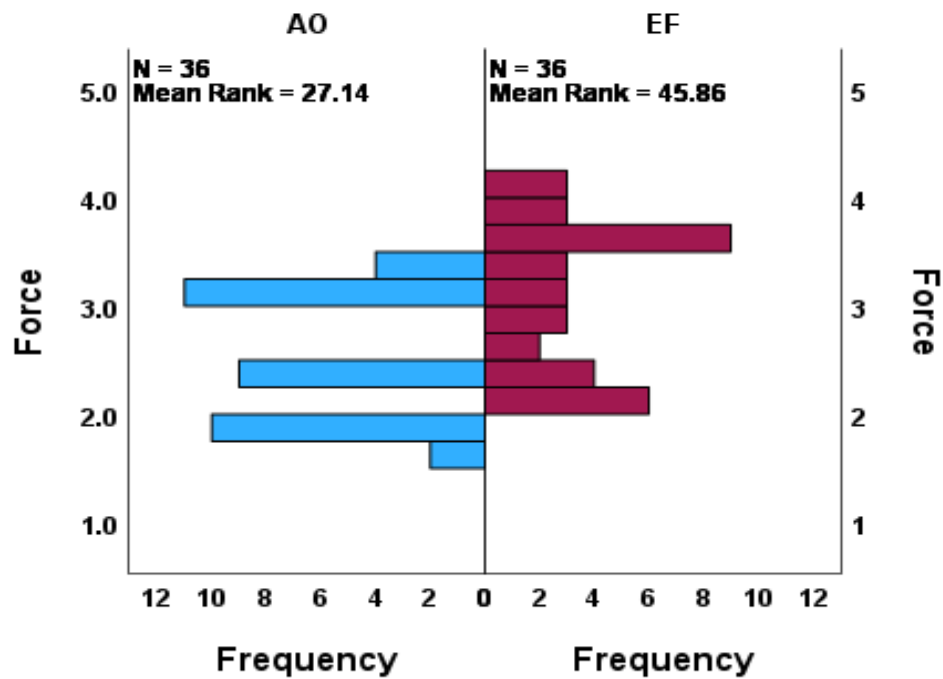


Figure 4.5. Comparison of the maximum force imparted by Euroform and American Orthodontics archwires, using Independent-Samples Mann-Whitney U Tests when archwires are displaced to 3 mm

**Table 4.12. The maximum, minimum and mean force (N) for 0.012” and 0.014” archwires (both brands, Euroform (EF) and American Orthodontics (AO) separated) at (a) maximum strain of 3 mm (b) the midpoint, taken at strain of 1.5 mm, of the unloading phase**

|           | Force (N) measured at:      |         |      |                                            |         |      |
|-----------|-----------------------------|---------|------|--------------------------------------------|---------|------|
|           | (a) Maximum strain of 3 mm: |         |      | (b) Midpoint of unloading phase at 1.5 mm: |         |      |
|           | Minimum                     | Maximum | Mean | Minimum                                    | Maximum | Mean |
| 0.012” EF | 2.3                         | 3.4     | 2.6  | 0.5                                        | 0.7     | 0.6  |
| 0.012” AO | 1.8                         | 2.4     | 2.2  | 0.0                                        | 0.3     | 0.1  |
| 0.014” EF | 3.3                         | 4.2     | 3.7  | 0.8                                        | 1.4     | 1.1  |
| 0.014” AO | 2.0                         | 3.5     | 3.1  | 0.3                                        | 0.6     | 0.5  |

#### **4.2.1.2.1 0.012” archwire measurements**

A wider range of variability in the Euroform group (2.3N – 3.4N) compared to the American Orthodontics group (1.8 – 2.4N) was noted in relation to the maximum force imparted by the archwires (Table 4.12 A & B). The archwires from both manufacturers appear to exhibit a similar loading plateau pattern. However, the force delivered by the American Orthodontics archwires (0.0N - 0.3N) during the unloading phase is broadly half that delivered by the Euroform wires (0.5N - 0.7N; Table 4.12 A & B) The minimum force imparted at the midpoint of the unloading curve for one archwire in the 0.012”

American Orthodontics group was recorded as 0.0N due to the 'jamming effect' noted during maximum displacement (Figure 4.7).

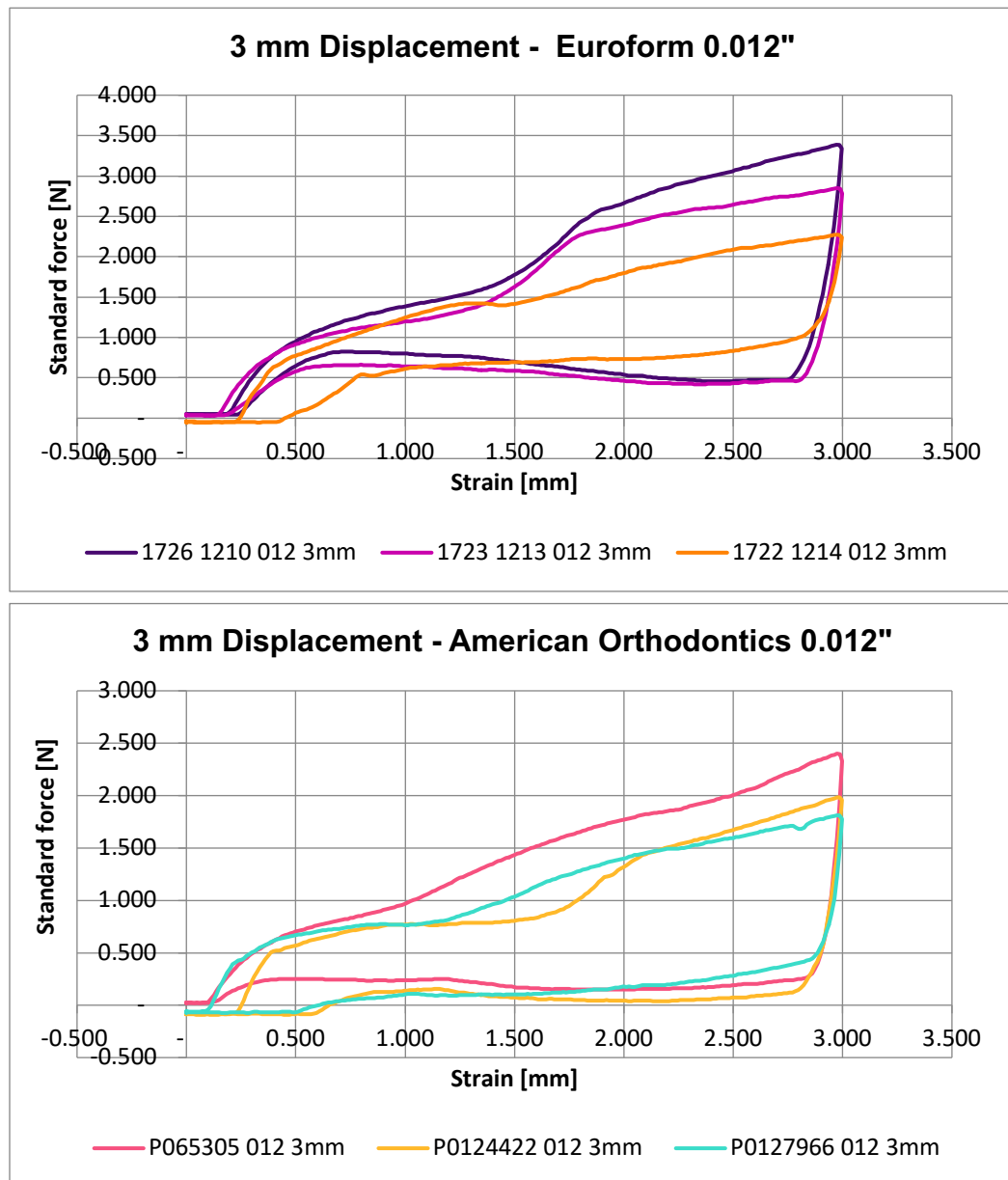


**Figure 4.6 A & B. Load deflection graphs showing the characteristics of 0.012" archwires from Euroform and American Orthodontics at maximum displacement of 3 mm**

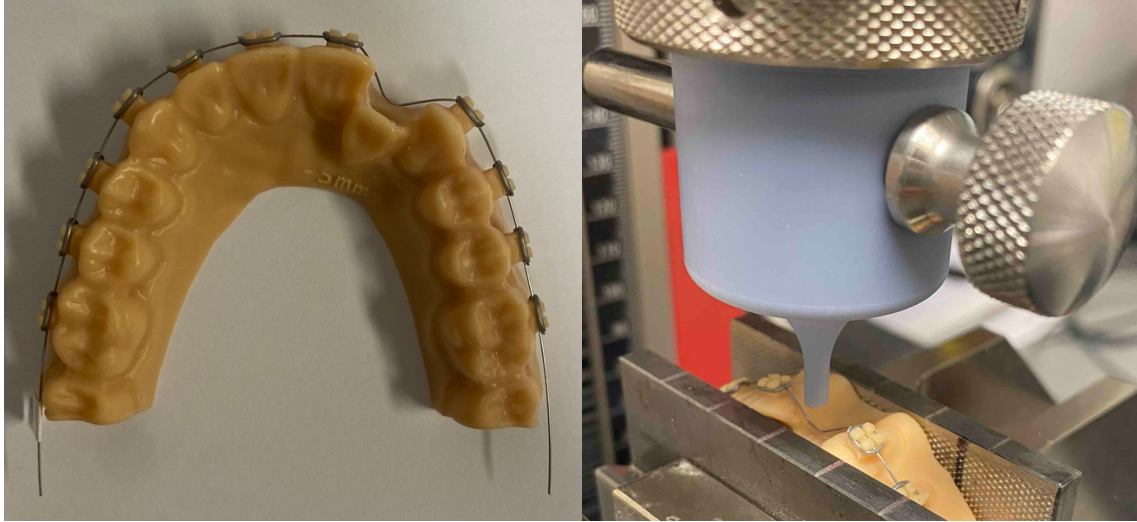


**Figure 4.7. ‘Jamming effect’ whereby the archwire does not return to its original form from maximum displacement**

#### **4.2.1.2.2 0.014” archwire measurements**

A wider range of variability in the American Orthodontics group (2N – 3.5N) was noted compared to the Euroform group (3.3N – 4.2N) in respect of the maximum force imparted by the archwires (Figure 4. A & B). A large difference in the maximum force imparted by wires from both manufacturers was observed with a maximum force of 4.2N by Euroform in comparison to a maximum force of 3.5N for American Orthodontics. The archwires from both manufacturers appear to follow a similar loading plateau pattern. However, again the force delivered by the American Orthodontics archwires (0.3N - 0.6N) during the unloading phase was almost half that delivered by the Euroform wires (0.8N – 1.4N; Table 4.12).

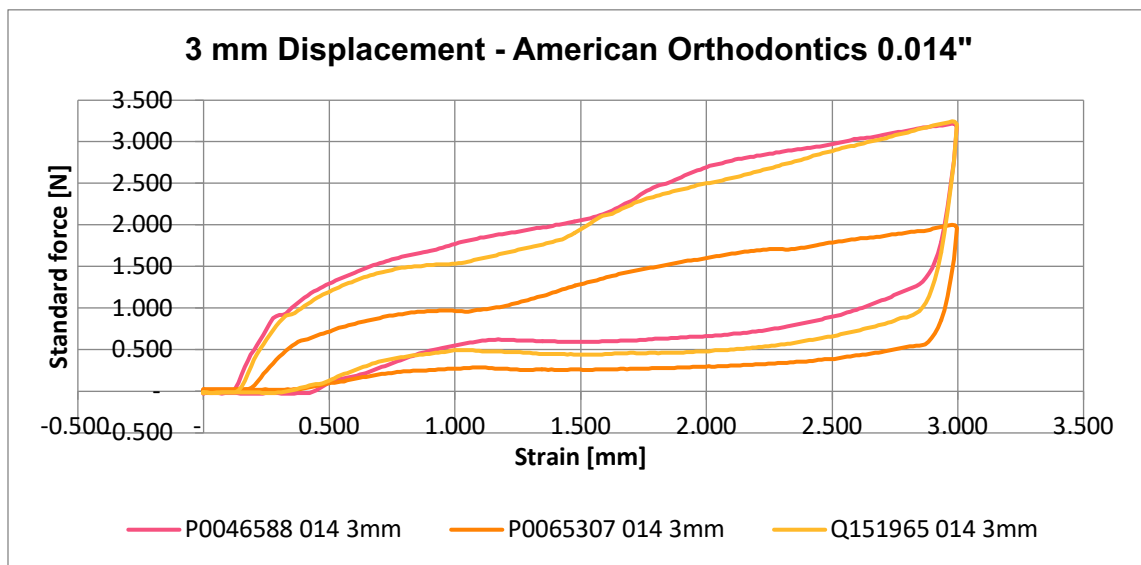
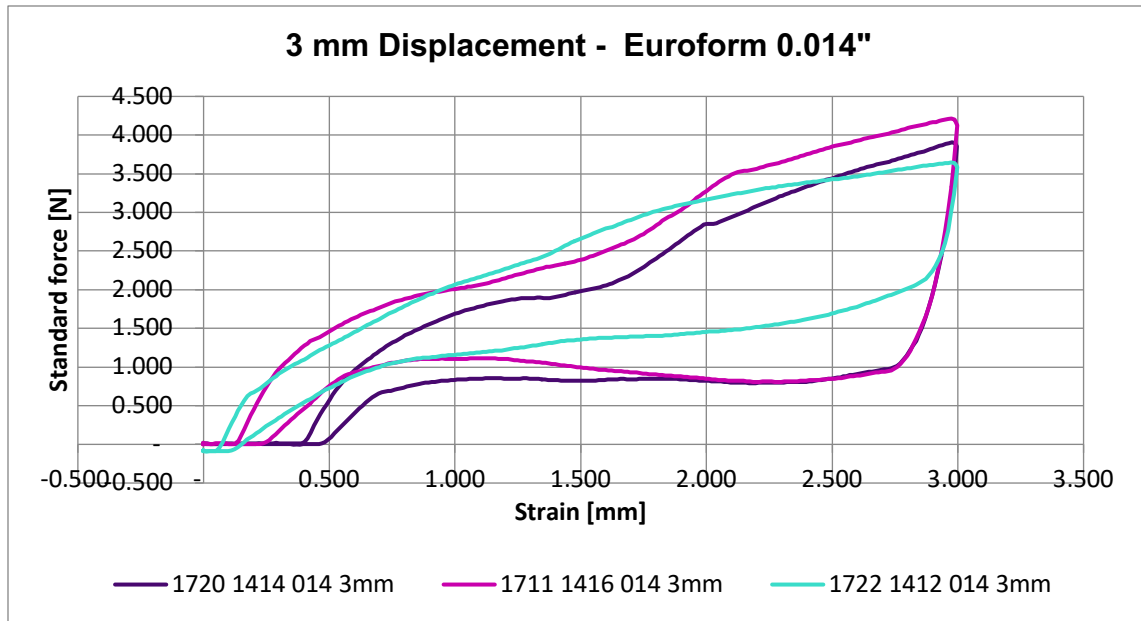


**Figure 4.8 A & B. Load deflection graphs showing the characteristics of 0.014" archwires from Euroform and American Orthodontics at maximum displacement of 3 mm**

#### 4.2.1.3 Effect of different batches within same manufacturers

For both Euroform and American Orthodontics archwires at 3 mm displacement, the difference between batches within each manufacturer was significant ( $P < .001$ ) (Table 4.13).

**Table 4.13. Statistical analysis of the maximum force imparted between all batches within each brand (Euroform and American Orthodontics) at archwire displacement of 3 mm**

| Brand                 | Max Strain | Chi-sq | df | P value |
|-----------------------|------------|--------|----|---------|
| American Orthodontics | 3 mm       | 24.79  | 5  | <.001   |
| Euroform              | 3 mm       | 25.37  | 5  | <.001   |

#### 4.2.2 Lateral incisor displacement of 4 mm

The loading gradient varied among the archwires with some showing a steeper gradient while others followed a more characteristic superelastic pattern. There was variation in the force delivered at maximum displacement.

#### **4.2.2.1 Effect of archwire cross sectional diameter (0.012" vs 0.014") on the maximum force**

The maximum force imparted by 0.012" and 0.014" nickel titanium archwires including both manufacturers (Euroform and American Orthodontics) at 4 mm displacement are presented in Figure 4.9 A & B. There appears to be a range of maximum forces imparted for both 0.012" (2.4N – 3.9N) and 0.014" (3.3N – 4.8N) archwires (Table 4.14).

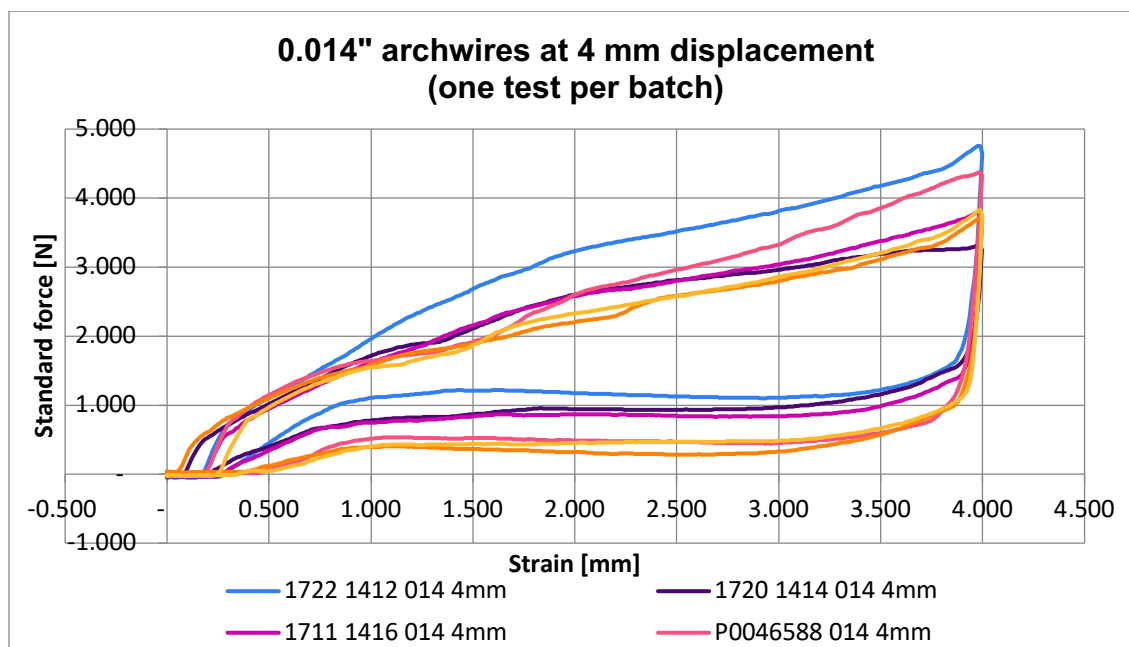
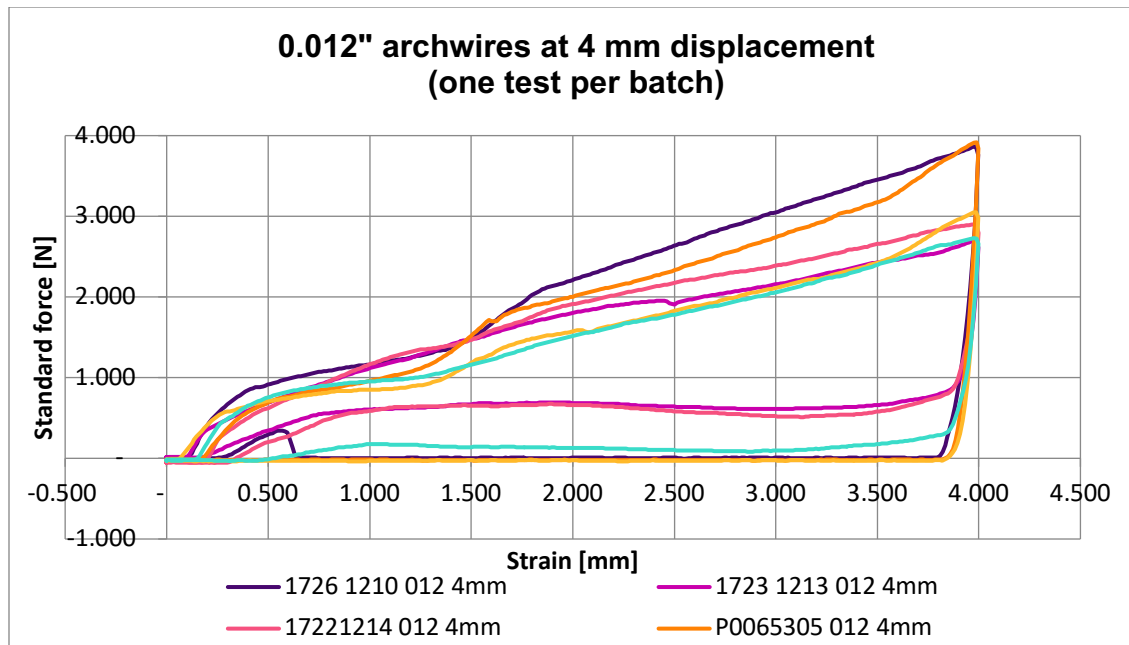


**Figure 4.9 A & B. Load deflection graphs showing the characteristics of 0.012" and 0.014" archwires from Euroform and American Orthodontics at maximum displacement of 4 mm**

**Table 4.14. The maximum and minimum force (N) for 0.012” and 0.014” archwires (both brands combined) at (a) maximum strain of 4 mm (b) the midpoint, taken at strain of 2 mm, of the unloading phase**

|        | Force (N) measured at:      |         |      |                                          |         |      |
|--------|-----------------------------|---------|------|------------------------------------------|---------|------|
|        | (a) Maximum strain of 4 mm: |         |      | (b) Midpoint of unloading phase at 2 mm: |         |      |
|        | Minimum                     | Maximum | Mean | Minimum                                  | Maximum | Mean |
| 0.012” | 2.4                         | 3.9     | 3.0  | 0.0                                      | 0.7     | 0.3  |
| 0.014” | 3.3                         | 4.8     | 4.0  | 0.3                                      | 1.3     | 0.7  |

#### **4.2.2.2 Effect of different archwire manufacturers**

No statistically significant difference in the maximum force imparted at 4 mm displacements between Euroform and American Orthodontics, when examined using Independent-Samples Mann-Whitney U Tests ( $P = 0.946$ ), was observed (Figure 4.10).

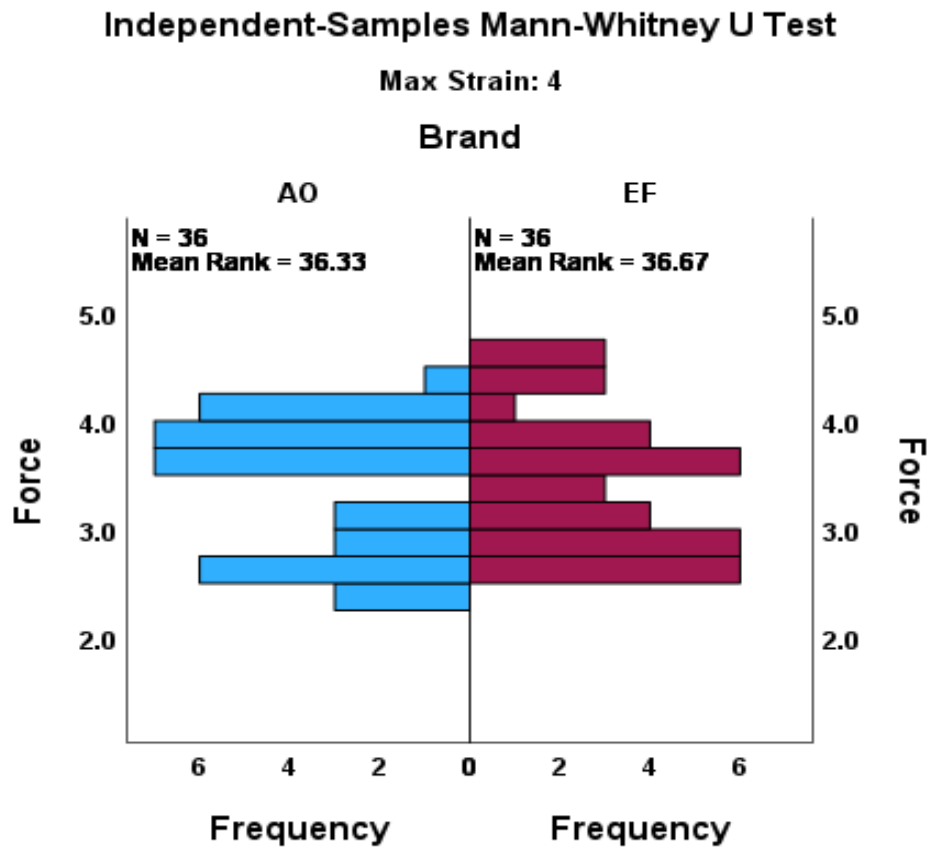


**Figure 4.10. Comparison of the maximum force imparted by Euroform and American Orthodontics archwires, using Independent-Samples Mann-Whitney U Tests when archwires are displaced to 4 mm**

**Table 4.15. The maximum, minimum and mean force (N) for 0.012” and 0.014” archwires (both brands, Euroform and American Orthodontics separated) at (a) maximum strain of 4 mm (b) the midpoint, taken at strain of 2 mm, of the unloading phase**

|           | Force (N) measured at:      |         |      |                                          |         |      |
|-----------|-----------------------------|---------|------|------------------------------------------|---------|------|
|           | (a) Maximum strain of 4 mm: |         |      | (b) Midpoint of unloading phase at 2 mm: |         |      |
|           | Minimum                     | Maximum | Mean | Minimum                                  | Maximum | Mean |
| 0.012” EF | 2.7                         | 3.9     | 3.1  | 0.0                                      | 0.7     | 0.5  |
| 0.012” AO | 2.4                         | 3.9     | 3.0  | 0.0                                      | 0.1     | 0.1  |
| 0.014” EF | 3.3                         | 4.8     | 4.0  | 0.8                                      | 1.3     | 1.1  |
| 0.014” AO | 3.7                         | 4.4     | 4.0  | 0.3                                      | 0.6     | 0.4  |

#### **4.2.2.2.1 Archwire dimension of 0.012”**

A similar range of variability in the American Orthodontics group (2.4N – 3.9N) compared to the Euroform group (2.7N – 3.9N) was noted in relation to the maximum force imparted by the archwires displaced to 4 mm (Figure 4.11 A & B). The archwires from both manufacturers appear to exhibit a similar loading plateau pattern. However, the force delivered by the American Orthodontics archwires (0.0N – 0.1N) during the unloading phase broadly half that delivered by the Euroform archwires (0.0N – 0.7N; Table 4.15) For both 0.012” archwire groups, the minimum force imparted at the midpoint of the unloading curve is recorded as 0.0N due to the ‘jamming effect’ noted during maximum displacement.

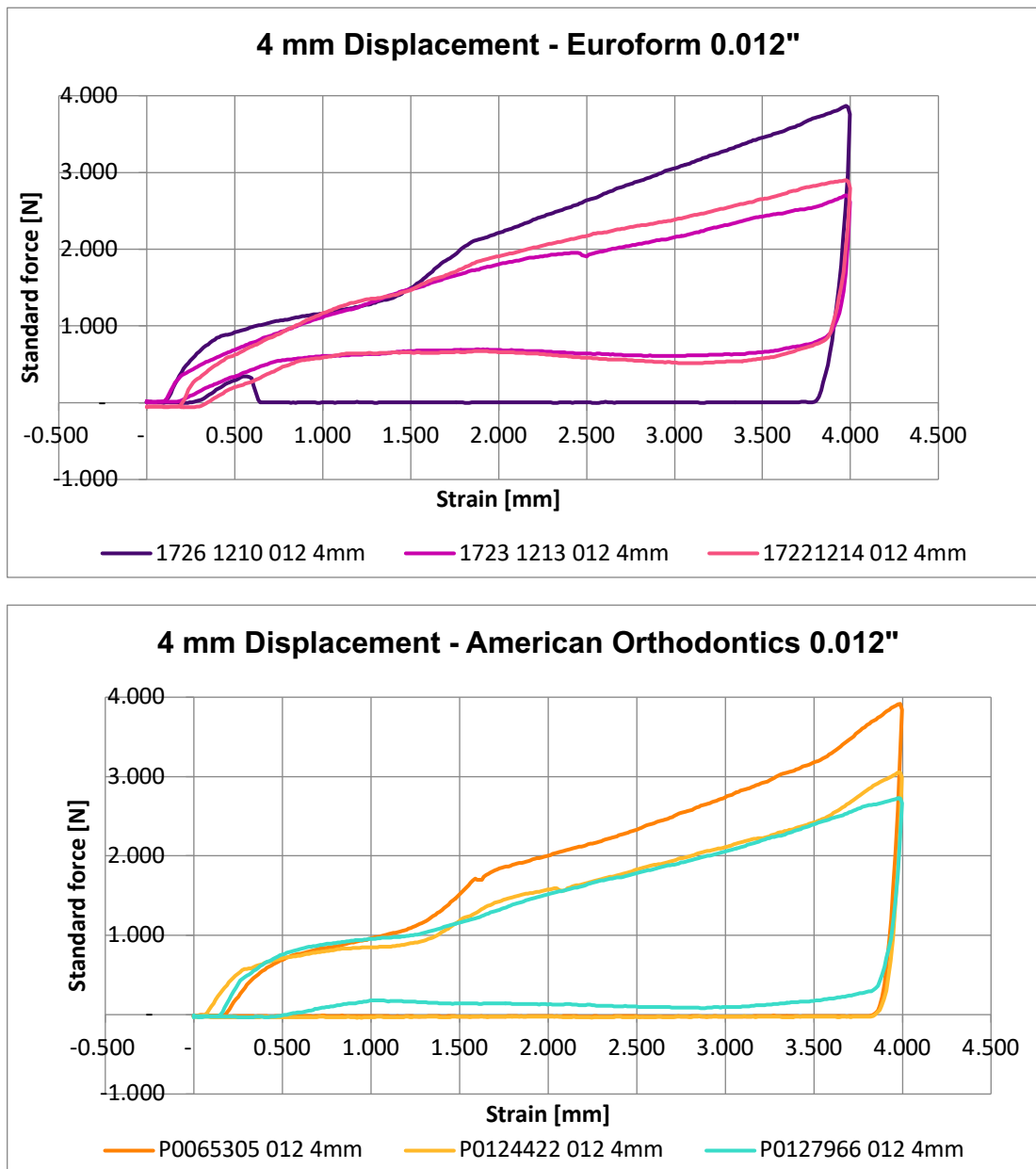


**Figure 4.11 A & B. Load deflection graphs showing the characteristics of 0.012" archwires from Euroform and American Orthodontics at maximum displacement of 4 mm**

#### **4.2.2.2.2 Archwire dimension of 0.014"**

A similar range of variability in the 0.014" American Orthodontics group (3.7N – 4.4N) compared to the Euroform group (3.8N – 4.8N) in respect of the maximum force imparted by the archwires displaced to 4 mm (Figure 4.12 A & B). The archwires from both manufacturers appear to follow a similar loading plateau pattern. However, the force delivered by the American Orthodontics archwires (0.3N – 0.6N) during the unloading phase was almost half that delivered by the Euroform wires (0.8N -1.3N; Table 4.15).

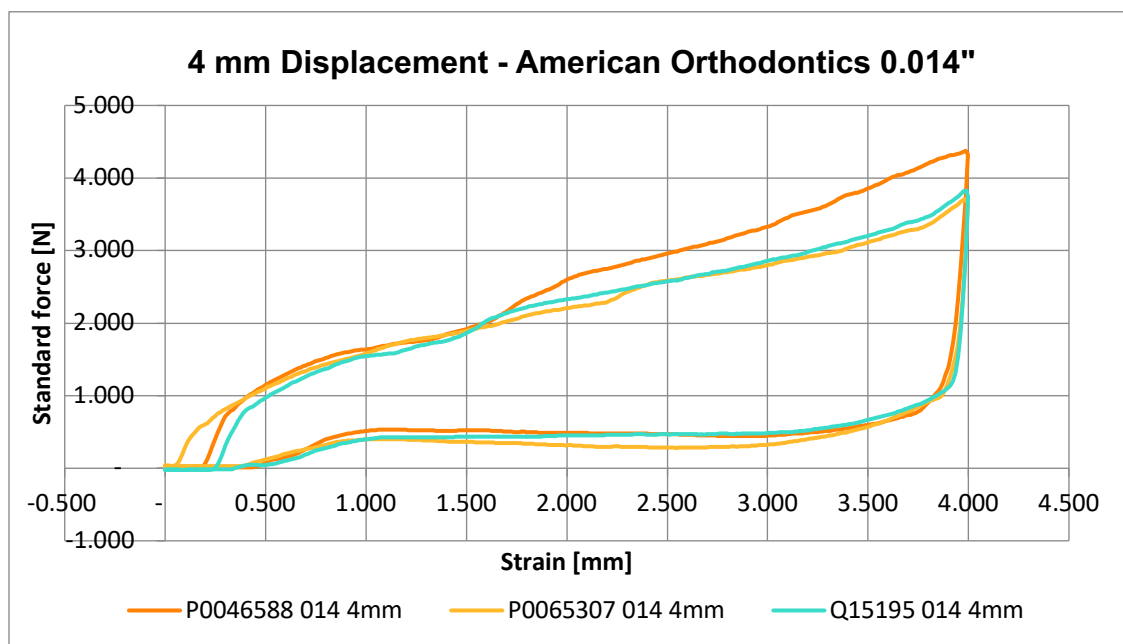
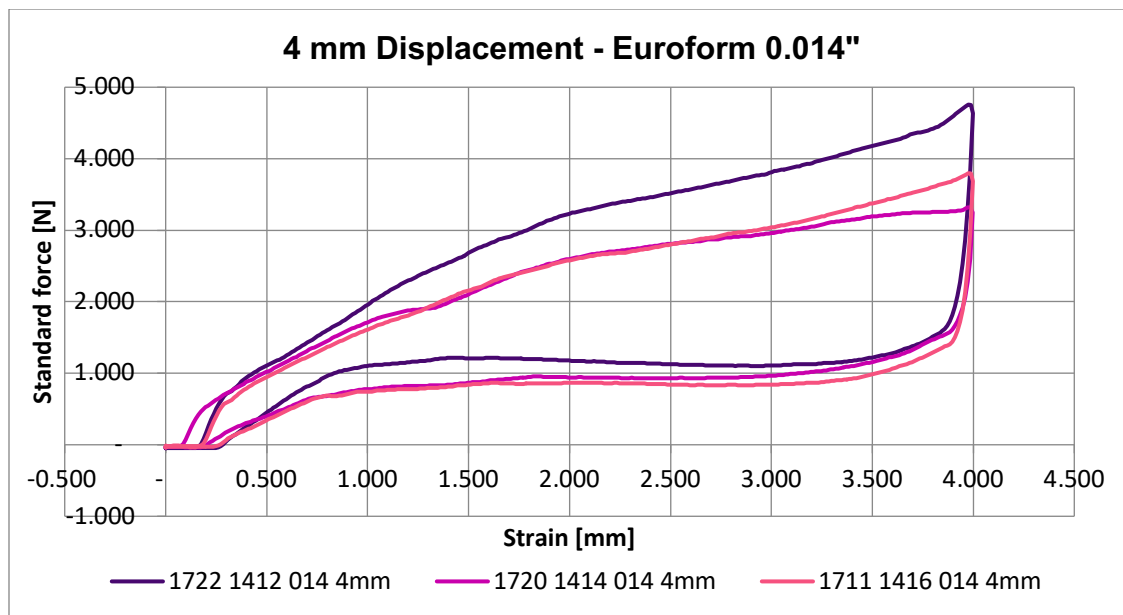


**Figure 4.12 A & B. Load deflection graphs showing the characteristics of 0.014" archwires from Euroform and American Orthodontics at maximum displacement of 4 mm**

#### 4.2.2.3 Effect of different batches within same manufacturers

For both Euroform and American Orthodontics archwires at 4 mm displacement, the difference between batches within each manufacturer was significant (Table 4.16).

**Table 4.16. Statistical analysis of the maximum force imparted between all batches within each brand (Euroform and American Orthodontics) at archwire displacement of 4 mm**

| Brand                 | Max Strain | Chi-square | df | p     |
|-----------------------|------------|------------|----|-------|
| American Orthodontics | 4 mm       | 25.36      | 5  | <.001 |
| Euroform              | 4 mm       | 26.35      | 5  | <.001 |

#### 4.2.3 Lateral incisor displacement of 5 mm

The loading gradient varied among the archwires with some showing a steeper gradient while others followed a more characteristic superelastic pattern. The force delivered at maximum displacement varied. Similar to the results of the 3 mm and 4 mm displacements, a 'jamming effect' was observed at 5 mm displacements in eight 0.012" archwires (3 x Euroform archwires and 5 x American Orthodontics archwires) and three 0.014" archwires (all American Orthodontics). The results for the unloading phase in the following section should be interpreted with caution. The force imparted by the archwire at maximum displacement and during the unloading phase

depending on the cross-sectional diameter, manufacturer, and batch of archwires used are considered.

#### **4.2.3.1 Effect of archwire cross sectional diameter (0.012" vs 0.014") on the maximum force**

The maximum force imparted by 0.012" and 0.014" nickel titanium archwires including both manufacturers (Euroform and American Orthodontics) at 5 mm displacements are presented in Figure 4.13 A & B. There appears to be a wide range in the difference in values of the maximum force imparted by both 0.012" (3.2N – 4.9N) and 0.014" (4.9N – 6.7N; Table 14.17).

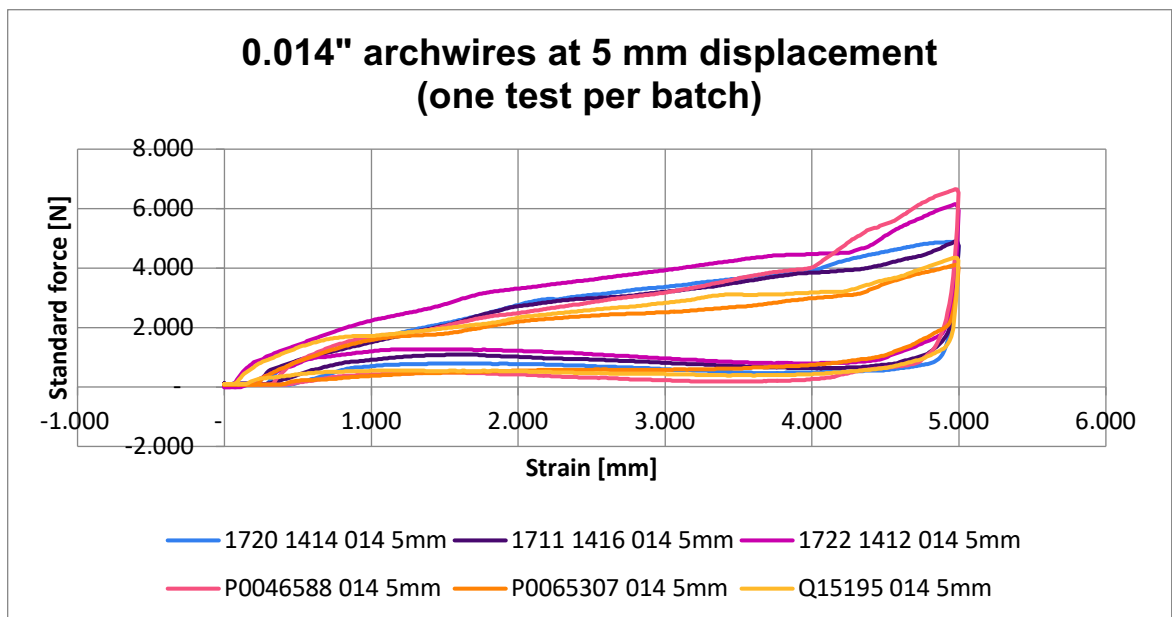
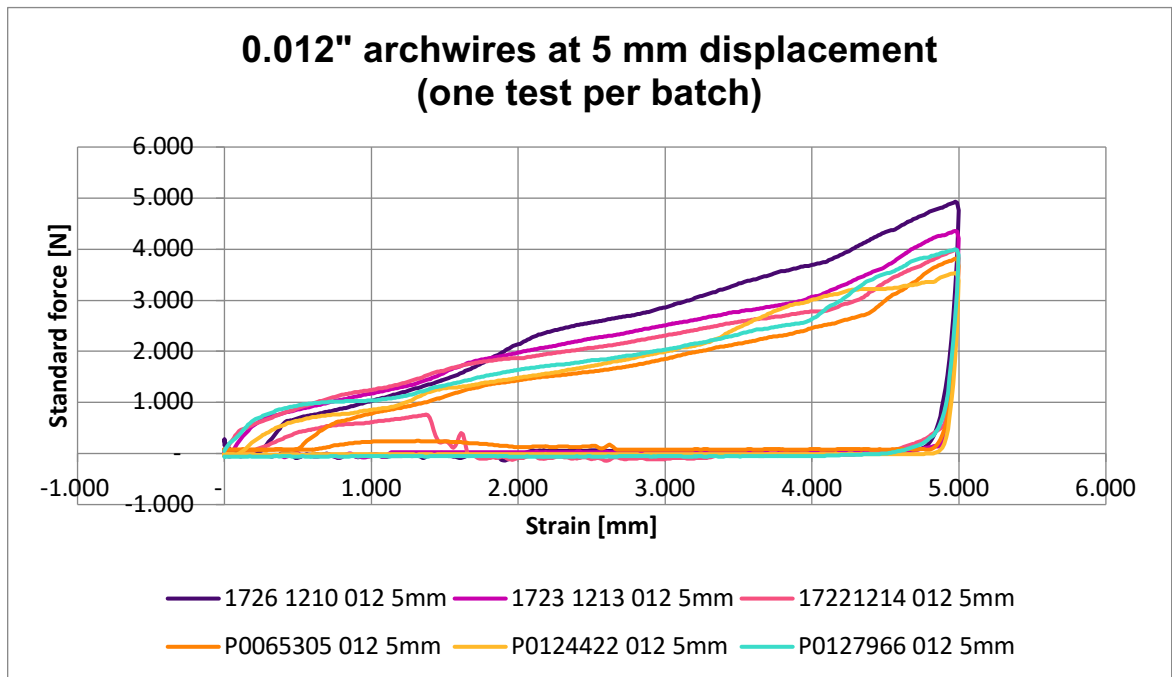


**Figure 4.13 A & B. Load deflection graphs showing the characteristics of 0.012" and 0.014" archwires from Euroform and American Orthodontics at maximum displacement of 5 mm**

**Table 4.17. The maximum and minimum force (N) for 0.012” and 0.014” archwires (both brands combined) at (a) maximum strain of 5 mm (b) the midpoint, taken at strain of 2.5 mm, of the unloading phase**

|        | Force (N) measured at:     |         |      |                                           |         |      |
|--------|----------------------------|---------|------|-------------------------------------------|---------|------|
|        | (a) Maximum strain of 5 mm |         |      | (b) Midpoint of unloading phase at 2.5 mm |         |      |
|        | Minimum                    | Maximum | Mean | Minimum                                   | Maximum | Mean |
| 0.012” | 3.2                        | 4.9     | 4.1  | 0.0                                       | 0.6     | 0.1  |
| 0.014” | 2.3                        | 6.7     | 5.2  | 0.0                                       | 1.1     | 0.6  |

#### **4.2.3.2 Effect of different archwire manufacturers**

A statistically significant difference in the maximum force imparted at 5 mm displacements between Euroform and American Orthodontics archwires when examined using Independent-Samples Mann-Whitney U Tests ( $P = 0.006$ ), was observed (Figure 4.15). Independent-Samples Mann-Whitney U Tests were used to determine if both archwire dimensions were significant within the brands and a statistically significant difference was observed for 0.012” archwires ( $P = 0.001$ ) while no statistically significant difference was observed for 0.014” archwires ( $P = 293$ ).

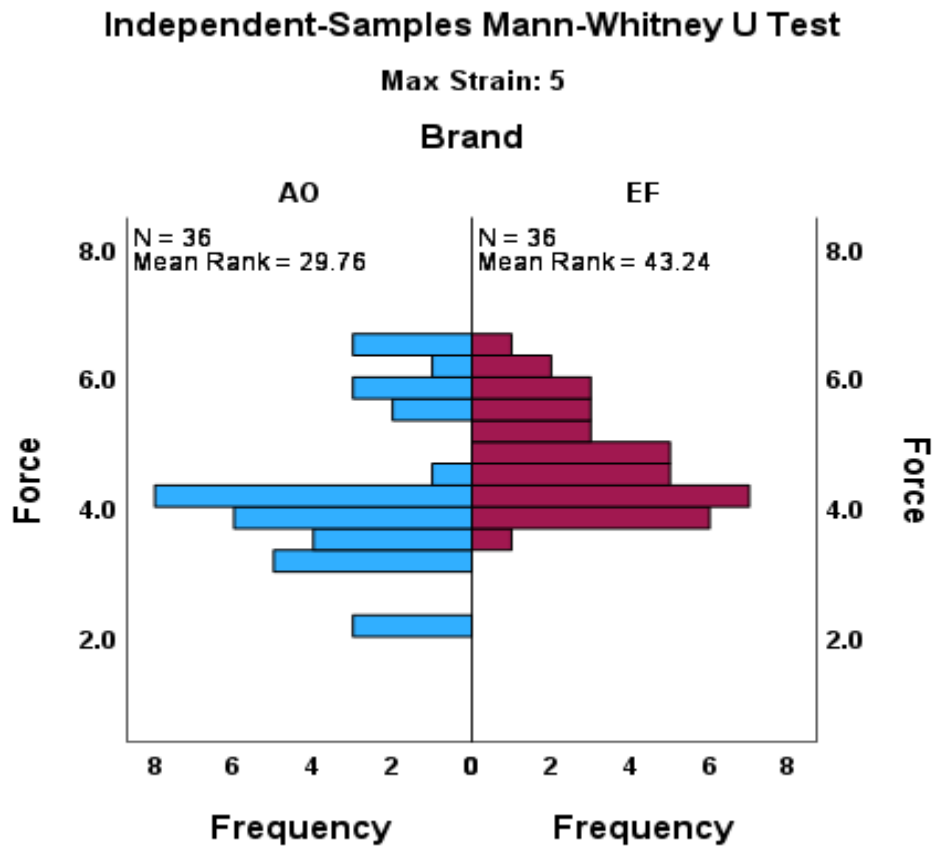


**Figure 4.14.** Comparison of the maximum force imparted by Euroform and American Orthodontics archwires, using Independent-Samples Mann-Whitney U Tests when archwires are displaced to 5 mm

**Table 4.18. The maximum, minimum and mean force (N) for 0.012” and 0.014” archwires (both brands, Euroform (EF) and American Orthodontics (AO) separated) at (a) maximum strain of 5 mm (b) the midpoint, taken at strain of 2.5 mm, of the unloading phase**

|           | Force (N) measured at:     |         |      |                                           |         |      |
|-----------|----------------------------|---------|------|-------------------------------------------|---------|------|
|           | (a) Maximum strain of 5 mm |         |      | (b) Midpoint of unloading phase at 2.5 mm |         |      |
|           | Minimum                    | Maximum | Mean | Minimum                                   | Maximum | Mean |
| 0.012” EF | 3.8                        | 4.9     | 4.3  | 0.0                                       | 0.6     | 0.3  |
| 0.012” AO | 3.2                        | 4.6     | 3.8  | 0.0                                       | 0.1     | 0.0  |
| 0.014” EF | 4.9                        | 6.5     | 5.6  | 0.7                                       | 1.1     | 0.9  |
| 0.014” AO | 2.3                        | 6.7     | 4.9  | 0.0                                       | 0.6     | 0.2  |

#### **4.2.3.2.1 Archwire dimension of 0.012”**

A similar range of variability in the American Orthodontics group (3.2N – 4.6N) compared to the Euroform group (3.8N – 4.9N) was noted in relation to the maximum force imparted by the archwires (Figure 4.15 A & B). The archwires from both manufacturers appear to follow a similar loading plateau pattern. However, the force delivered by the American Orthodontics archwires (0.1N) during the unloading phase is much less than that delivered by the Euroform archwires (0.6N; Table 4.18). For both 0.012” archwire groups, the minimum force imparted at the midpoint of the unloading curve is recorded as 0.0N. This is due to the ‘jamming effect’ noted during maximum displacement and is

demonstrated on the load deflection diagram Figure 4.15 A & B. Due to the 'jamming effect' it is not possible to draw conclusions from these results.

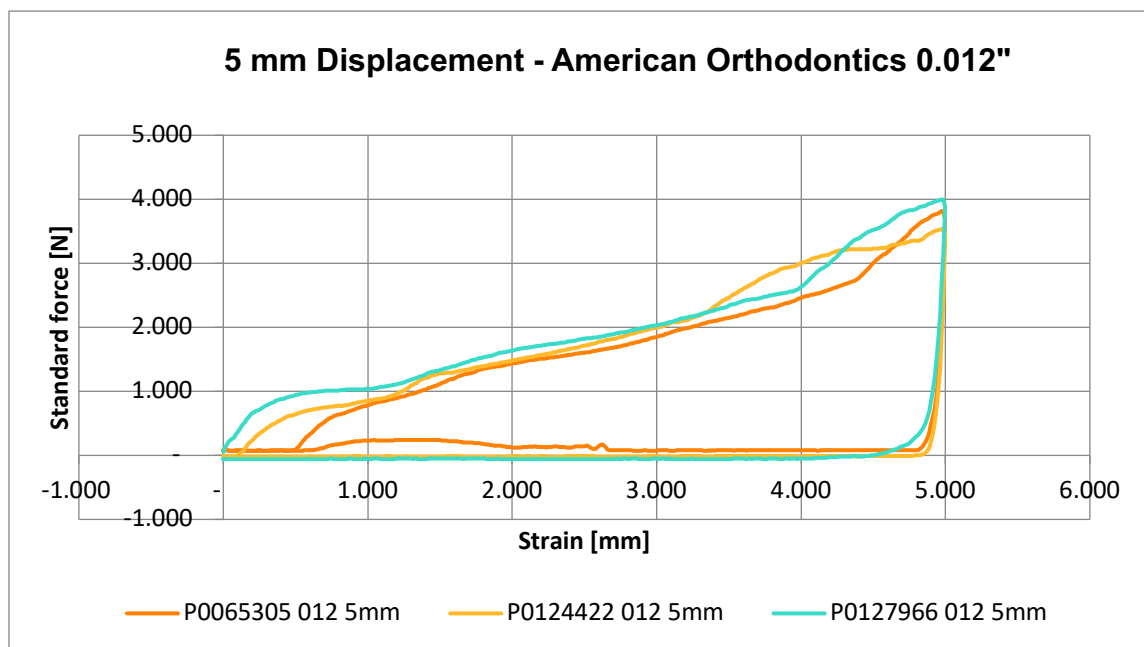
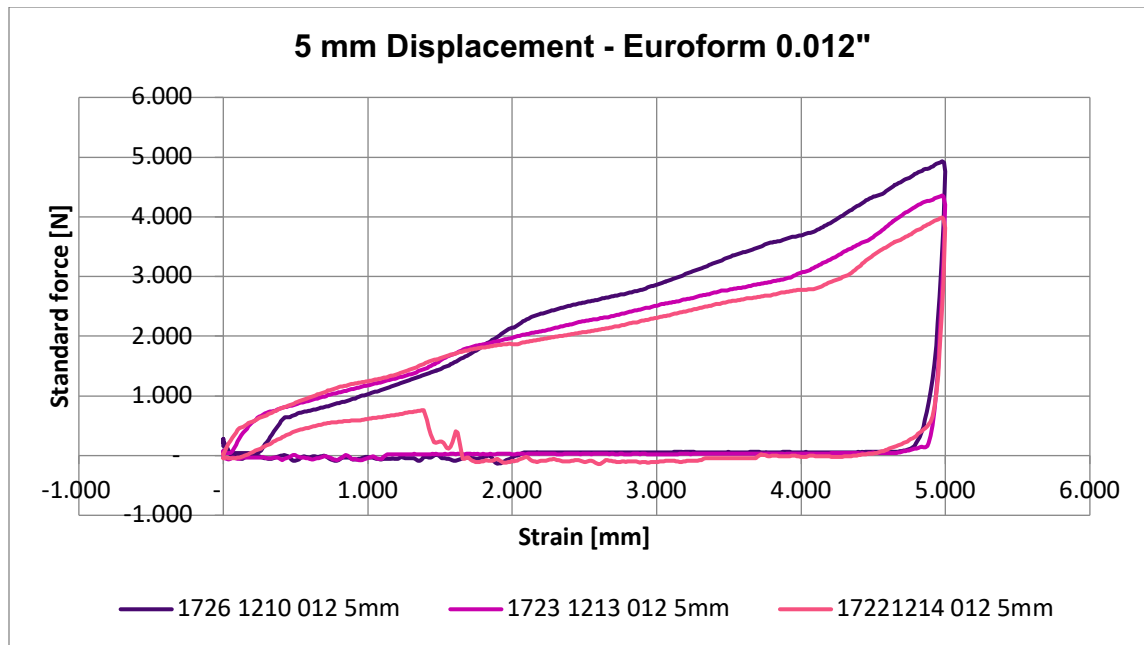


**Figure 4.15 A & B. Load deflection graphs showing the characteristics of 0.012" archwires from Euroform and American Orthodontics at maximum displacement of 5 mm**

#### **4.2.3.2.2 Archwire dimension of 0.014"**

A wide range of variability in the American Orthodontics group (2.3N – 6.7N) was noted compared to the Euroform group (4.9N – 6.5N) in respect of the maximum force imparted by the archwires (Figure 4.16 A & B). The archwires from both manufacturers appear to follow a similar loading plateau pattern. However, the force delivered by the American Orthodontics archwires (0.0N – 0.6N) during the unloading phase is broadly half that delivered by the Euroform wires (0.7N - 1.1N; Table 4.18). The minimum force recorded at the midpoint of the unloading curve for 0.014" American Orthodontics archwires was 0.0N due to the 'jamming effect' noted during maximum displacement.

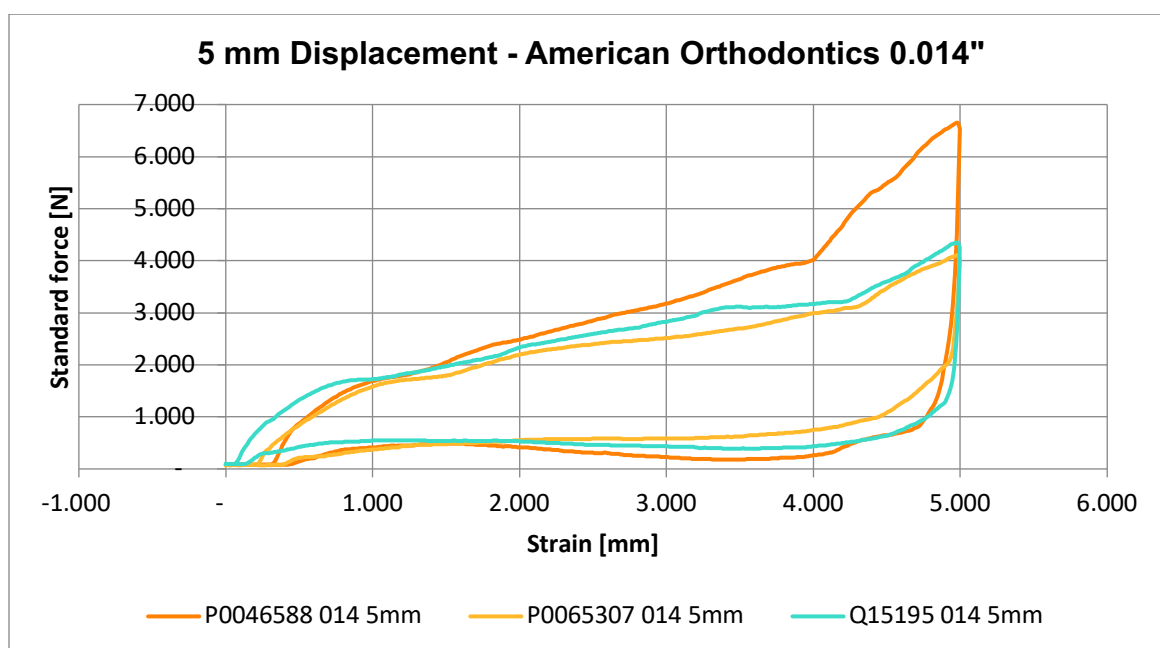
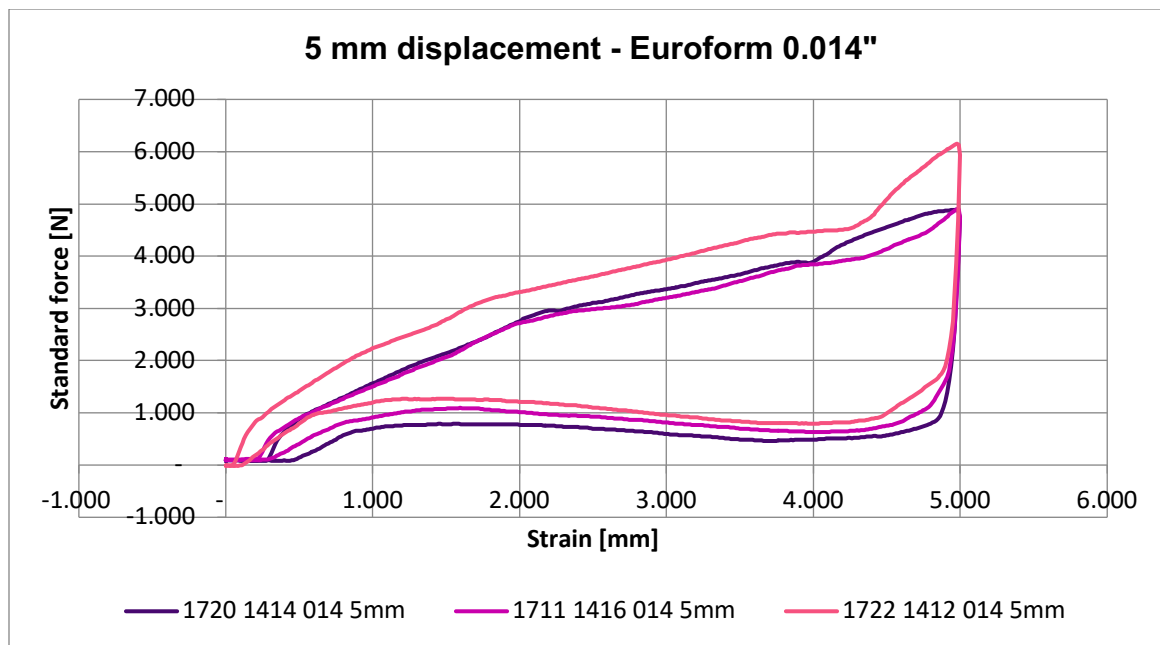


**Figure 4.16 A & B. Load deflection graphs showing the characteristics of 0.014" archwires from Euroform and American Orthodontics at maximum displacement of 5 mm**

#### 4.2.3.3 Effect of different batches within same manufacturers

For both Euroform and American Orthodontics archwires at 5 mm displacement, the difference between batches within each manufacturer was significant (Table 4.19).

**Table 4.19. Statistical analysis of the maximum force imparted between all batches within each brand (Euroform and American Orthodontics) at archwire displacement of 5 mm**

| Brand                 | Max Strain | Chi-square | df | p     |
|-----------------------|------------|------------|----|-------|
| American Orthodontics | 5          | 16.11      | 5  | 0.007 |
| Euroform              | 5          | 29.81      | 5  | <.001 |

#### 4.2.4 Overall comparison of the unloading phases

Looking at the overall unloading curves of the load deflection graphs, there was variability in both the force imparted and in terms of the 'slope' of the curve. While the above section discussed the unloading force at the midpoint of the unloading curve for each displacement category, the following section looks at the 'slope' of the unloading curve. Ideally, the force on the tooth should be continuous and low, represented by an almost horizontal unloading plateau at a low force level. The regression slope of the unloading curve between strain of 1 mm and 2 mm was analysed for each graph. The slope of each archwire, when compared to a slope of 0 i.e. a horizontal line, was

observed to be steady ranging between -0.10N to 0.09N. For two batches the slope was not significantly different to 0 ( $P > 0.05$ ), meaning a very steady force was imparted by the archwire. Two batches, however, showed a higher variation in force between strain of 2mm to 1mm in the unloading phase (Figure 4.17 A & B).

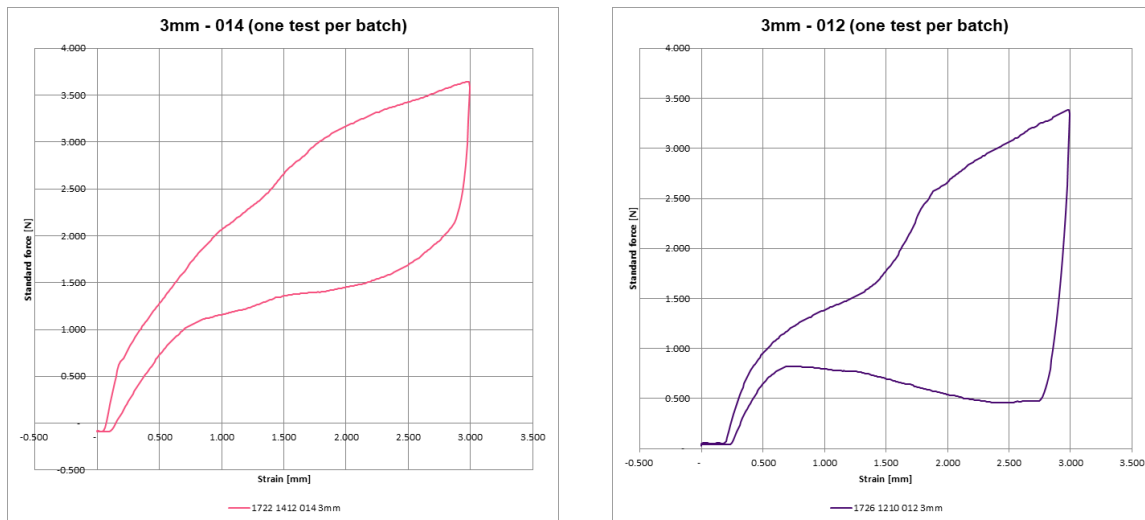


**Figure 4.17 A & B. Examples of large variations in unloading plateaus at strain between 1 mm – 2 mm**

## **Chapter 5 – Discussion**

The main purpose of this study was to compare the frequency and extent of permanent deformation of superelastic nickel titanium aligning archwires from the same and different manufacturers when engaged in the brackets of teeth of varying displacements in a simulated set up. A secondary purpose was to measure the force imparted onto a tooth as the degree of displacement increased from 3 mm to 5 mm.

### **5.1 Discussion of methodology**

This study aimed to replicate clinical scenarios as much as possible. Many testing methods have been discussed previously in the literature including brackets attached to acrylic blocks of varying designs (Lombardo *et al.*, 2012), moveable acrylic rods (Higa *et al.*, 2017) and typodont teeth (Andreasen & Barrett, 1973; Barrowes, 1982; Hudgins *et al.*, 1990). A palatally displaced lateral incisor tooth was simulated as this is a common presentation reflecting underlying dento-alveolar disproportion and the local eruption pattern. Various treatment approaches can be employed to correct this, including the use of elastomeric auxiliaries, piggyback mechanics or continuous archwire mechanics. Both of the latter involve a flexible, small-dimension nickel titanium archwire ligated to the displaced tooth to achieve alignment.

3D-printed models, which had been designed digitally, were used in this study to provide a reproducible testing set-up. Eight models were created for each increment of lateral incisor displacement which increased testing efficiency and reduced the wear and tear of the models. Not only did the printing of eight

identical 3D models of each increment allow eight archwires to be tested simultaneously, but it also allowed for identical testing conditions when examining the properties of the archwires. The 3D-printed models were printed with Gemini™ (3m Unitek) brackets templates attached to the teeth. Again, this improved the accuracy of the testing models as probable human error in bracket placement was removed. The inter-bracket distance used was representative approximating those deployed by Wilkinson *et al.* (2002) and Nucera *et al.* (2014) based on the average tooth dimensions for a male's maxillary dentition (Moyers, 1976).

## **5.2 Archwire selection and testing**

Two brands of archwires were selected, Euroform and American Orthodontics. These were chosen as they are used extensively in orthodontic treatment. The mechanical properties of 0.016" nickel titanium archwires have previously been tested (Bellini *et al.*, 2016). Two smaller archwire dimensions, 0.012" and 0.014", were chosen as they are the more suited to accommodating large archwire deflections during the initial alignment phase of orthodontic treatment. ISO 15841 recommends the evaluation of six archwire specimens from each batch of archwires. Higa *et al.* (2017) used 10 archwires per batch to minimise the chance of any technical errors that could influence the results. In order to allow a control archwire during Part I of the study and allow two archwires for testing in Part II, eight archwires were chosen for Part I. This was deemed sufficient due to the controlled nature of the laboratory-based study meaning a large sample size was not required. Typically, 3-point wire bending

tests are carried out on a straight portion of an archwire not confined within a fixed appliance (Bellini *et al.*, 2016; Nakano *et al.*, 1999). In this study, 3-point bending tests were completed on the curved portion of the wire at the maxillary lateral incisor site while the archwire was ligated to brackets of adjacent teeth. It was felt again that this set-up better replicates clinical scenarios. Adaptations were required to the laboratory set-up in order to permit this.

### **5.3 Quantification of archwire deformation**

While previous studies have mostly tested archwires through 3-point bending tests, assessment of whether an archwire is clinically usable has not been made (Bellini *et al.*, 2016; Brauchli *et al.*, 2011; Nakano *et al.*, 1999). Significant tooth displacement may dictate re-ligation at the next appointment if it remains active. With this in mind, when a nickel titanium archwire is removed from a dental arch it can return to the original archform as expected, given its properties, or it can be permanently deformed. The issue with permanent deformation is that it is impossible to know at what stage the deformation occurred, i.e. initially after ligation or later. If it occurs early, the archwire may lack potency for the duration of the inter-appointment interval meaning little to no tooth movement would occur. This results in longer and more inefficient orthodontic treatment. A time period of 24 hours was chosen in this study to examine the initial deformation that could occur when the archwires are deflected considerable distances. This observation period was informed by

previous research whereby similar deformation of the archwire was noted at 24 hours versus 28 days (Hudgins *et al.*, 1990).

Development and validation of an index allowed the used archwires to be categorised according to their condition. The objective deformation parameters used in this study were angular and linear measurements. The archwires were placed on a flat surface for all measurement to be taken. All archwires, except for two, could be placed flat on the photocopy of the unused wire meaning they were deformed in two planes rather than three. This allowed accurate angular and linear measurements to be made. The most distal point of the used wire was used as the measurement point as it was easily and accurately identifiable and allowed comparison between the archwires. The position of the angular measurement along the archwire varied meaning that direct comparison between wires was not entirely accurate, however the measurement still provided information on the extent of the deformation.

A comparison of outcomes from the subjective (i.e. index) and objective (i.e. linear and angular) measurements was performed. An index category of 1 (no deformation) typically corresponded to linear measurements below 1 mm and angular measurements below 7.5°. An index category of 2 (minor deformation) typically corresponded to linear measurements below 4 mm and angular measurements below 12.5°, where at least one of the index category 1 thresholds was breached. An index category of 3 (major deformation) typically applied where at least one of the index category 2 thresholds was breached.

In previous research, the archwire was released from the displaced tooth only and the degree of deformation was measured as a percentage of the distance from the bracket to the releases archwire versus the same distance to the unused archwire before ligation (Andreasen & Hilleman, 1971). In this study the most displaced tooth was released first with consideration given to including this test as a measure of deformation. However, it was noted that occasionally the archwire did not return completely to its original form while still being ligated to the adjacent teeth suggesting that there was deformation, but in fact, once the archwire was fully released from the typodont, it often returned to the original form.

#### **5.4 Evaluation of the unloading plateau**

Brauchli *et al.* (2011) compared the properties of three types of orthodontic wires and measured the force imparted by the wire during the unloading phase. The force value was taken at the midpoint of the unloading curve (1.5 mm as the wire was deflected to 3 mm), allowing comparison between the wire types and brands. This concept was used in this study when looking at the unloading plateaus for the displacements of 3-, 4- and 5- mm. A ‘jamming effect’ was noted when displacing archwires to a distance of 3 mm, 4 mm or 5 mm in 15 out of 72 archwires tested (Figure 4.7). This meant that the archwire was deflected to 4 or 5 mm by the load cell but did not return with the cell as the load was removed and remained static. It was interesting to note that the 0.012” archwires taken from American Orthodontics batch number P0124422 experienced the ‘jamming effect’ across 3 mm, 4 mm and

5 mm displacements while 0.012" archwires taken from American Orthodontics batch number P0124422 experienced the 'jamming effect' across 4 mm and 5 mm displacements. This effect was more prevalent with lighter 0.012" archwires. Out of the 15 archwires which experienced the 'jamming effect', 12 were 0.012" archwires across 3-, 4- or 5- mm displacements while the remaining 3 were 0.014" archwires displaced to 5 mm. The ability to compare the unloading plateau (the mean force, midpoint values or the slope) of the archwires and thus the force imparted onto a tooth was affected accordingly. This anomaly has also been reported by Elayyan *et al.* (2010) when examining the mechanical properties of coated and uncoated archwires observing that the unloading force values were zero at 4 mm displacement points due to the archwires 'jamming' at their most displaced point. Of note, the wire had not become permanently deformed in this 'jammed' state and returned to its original form once it was manually released. The effect is, therefore, unlikely to occur in a clinical scenario as occlusal or jiggling forces are likely to release the wire from this state. However, given that the effect is most likely to occur when using American Orthodontics 0.012" archwires at 5 mm displacements, it is something that could be taken into consideration clinically when choosing an archwire.

## **5.5 Quantification of archwire force**

Higa *et al.* (2017) noted that increasing in the archwire diameter increases the force imparted, i.e. 0.016" archwires exert a greater force than 0.014"

archwires. In this study it was deemed important to determine the force difference between 0.012" and 0.014" archwires at varying degrees of displacement. Higa *et al.* (2017) found that the mean force of 0.014" conventional nickel titanium archwires at a 3 mm deflection in a conventional bracket with elastomeric ligation was 2.77N. This is comparable to the current study whereby the mean maximum force for 0.014" archwires, regardless of the brand, was 3.4N. The mean maximum force of 3.1N exerted by American Orthodontics archwires at 3 mm displacement approximated the findings of Higa *et al.* (2017).

The force imparted on the displaced tooth was measured independently of the ligation method in this study. The archwire was deflected using a custom-made silicone trough to accommodate the wire while adjacent teeth were ligated with elastomeric modules. This is similar to the study by Lombardo *et al.* (2012), who used a metal blade to deflect the archwire to 4 mm and noted a 50% increase in the average plateau of the unloading force as archwires increased in diameter from 0.012" to 0.014". In terms of the pattern of the unloading force in the current study, a slight increase in force was seen at the initial stage of the unloading force (Figure 5.1). This was also noted by Lombardo *et al.* (2012) and was thought to be related to different friction coefficients between the wire and the brackets.

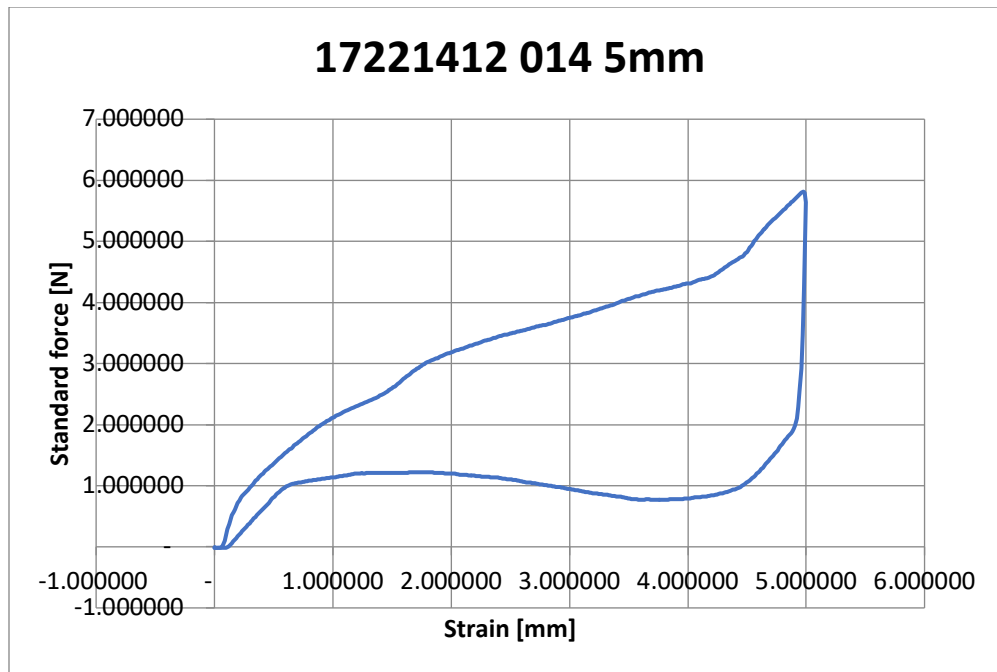


**Figure 5.1. Force deflection curve showing increase in force during the unloading plateau**

## **5.6 Comparing Euroform archwires and American Orthodontics archwires at displacements of 3-, 4- or 5- mm**

Despite the various tests being performed in a controlled environment with specific consistent displacement distances and, in the case of the typodont deformation tests, specific displacement settings and a fixed temperature, there exists significant variation in the categorical, linear and angular deformation of the archwires. There was also variation in the maximum force applied and the force measured during the unloading phase across brands, between archwire size (0.012" versus 0.014"), within brands and/or archwire size between batches and finally within brands.

### **5.6.1 Deformation of archwires**

When comparing the archwires falling within deformation categories 1, 2 or 3, Euroform archwires had a significantly greater number of archwires in the lower categories (1 or 2). Clinically, Euroform wires, regardless of the dimension (0.012" or 0.0014"), may be less likely to be deformed to the extent that renders them clinically unusable compared to American Orthodontics archwires.

Intuitively, one might expect that the deformation of a wire would increase as the displacement distance increases from 3 mm to 4 mm to 5 mm. This is supported by the linear measurement results seen for Euroform 0.012" archwires between displacements of 3 mm and 5 mm and between 4 mm and 5 mm. Similar results are seen for American Orthodontics 0.014" archwires

between 3 mm and 5 mm, with a broadly similar outcome for the angular measurements.

### **5.6.2 Force imparted by the archwire**

The force at maximum strain was significantly greater for 0.012" and 0.014" Euroform archwires compared to the same diameter American Orthodontics archwires when displaced 3 mm. The same was true when comparing the 0.012" archwires at 5 mm displacements. Clinically, American Orthodontics archwires may impart a lesser force onto the tooth at 3 mm to 5 mm displacements, which might be considered biomechanically preferable. When comparing 0.012" with 0.014" archwires, regardless of the brands, the force measured at maximum strain and at the midpoint of the unloading plateau is greater for 0.014" archwires resulting in an increased force imparted on the tooth. For this reason, clinically it may be better to use a 0.012" archwire for tooth displacements of 3 mm or more.

Based on the characteristics of the unloading plateau noted on the graphs as well as the slope value, force values appear more stable for American Orthodontics archwires in imparting a steady continuous force. Figure 4.18 shows the characteristics of two Euroform archwires showing large variations in the unloading plateau. Overall, the unloading comparisons cannot be used definitively due to the 'jamming effect' and the reduced sample size. Further research is needed to fully explore the unloading forces.

When looking at the maximum force between batches of the same brand, there were statistically significant differences in the maximum force imparted

by the archwire between the batches of the same brand at each displacement (3-, 4- or 5- mm). This is not ideal as it shows wide variation in the archwires supplied by a particular manufacturer. This result is in agreement with the research carried out by Bellini *et al.* (2016) and Nakano *et al.* (1999).

## **5.8 Limitations**

In terms of study design, this was a laboratory-based study; therefore, it did not fully replicate how the archwires could act clinically. Nevertheless, it would not be possible to replicate this study clinically as it would be impossible to standardise the deflection increments in patients and thus test the wires in an identical set-up. Nickel titanium archwires from two manufacturers were included in this study. Although these are the most commonly used brands within our clinical settings, it is recognised that many other comparable brands exist and many of them may exhibit unique properties.

Other factors that can influence tooth movement include the inter-bracket distance, type of bracket used, mode of ligation and the bracket-wire interface. These were not individually examined within this study. However, results from previous studies were taken into account to allow for comparison. For example, the inter-bracket distances chosen were those of Moyers *et al.* (1976) and were also used in the study carried out by Hudgins *et al.* (1990) and Nucera *et al.* (2014), while elastomeric modules were chosen. It is recognised that there is more friction at the wire-bracket interface with elastomeric modules but these were selected to avoid inadvertent deformation to the wire

during the ligation process. There was also potentially more friction at the wire-bracket interface that would have been the case if stainless-steel brackets were used.

In terms of materials used, tie wing fractures were noted on some of the 3D models (FotoDent® model2 385 nm) with these unable to withstand the force exerted by 0.014" archwires at 5 mm displacements. This may relate to the thickness of the material at the tie wings, a weakness in the material if unsupported during the 3D printing manufacturing process and the increased force exerted by the larger diameter archwires at higher deflections as seen in the force deflection graphs. Specifically, the maximum force exerted by 0.014" Euroform and American Orthodontics archwires was found to be 6.5N and 6.7N, respectively, during 3-point bending tests to a displacement of 5 mm. To facilitate completion of all planned testing, the remnants of the printed bracket were removed using an acrylic bur. A conventional metal Gemini™ (3M Unitek) bracket was then applied to the tooth using epoxy resin. It was possible to place this in the same position using the outline of the removed resin bracket.

In terms of the measurement of deformation, angular and linear measurements were used as the objective distortion parameters. However angular measurements were taken at different points along the archwires depending on where the deformation occurred. This complicates the comparison of the angular deformation across batches, brands and archwire

sizes. These findings were therefore coupled with the complementary measures to give a reliable measure of deformation.

## **5.9 Proposed future work**

During the alignment phase of orthodontic treatment, the same archwire is often reused and reactivated over a number of visits if teeth are quite displaced. It would be useful to know how much deflection an archwire can withstand or if it is more prone to deformation on its second or third reactivation. This would involve retesting the same archwires again for deformation using the same method as discussed above.

Archwires from two brands were included in this study. A future study could compare more archwire brands. In order to select commonly used brands among orthodontists, a survey among orthodontists could be carried out. This would improve the generalisability of the results.

Ideally, this project would be translated to a clinical study to overcome issues such as the 'jamming' effect and also to determine the effect that saliva and other intra oral factors have on archwire behaviour. However, due to variation in each person's malocclusion, standardisation of wire displacements would be impossible. This however could possibly be overcome through randomising the sample.

## **Chapter 6 - Conclusions**

Based on this experimental set-up, the use of an 0.012" archwire for tooth displacements of 3 mm or more may be considered in order to permit the delivery of optimal forces, due to the greater force imparted at maximum displacement by 0.014" archwires.. Additionally, the mean force produced by the American Orthodontics archwires was significantly less than for the Euroform archwires at the same magnitudes of displacement for both 0.012" and 0.014" sizes.

Euroform archwires performed better with less deformation relative to American Orthodontic wires. With this in mind, Euroform archwires could be less likely than American Orthodontics archwires to undergo a high level of permanent deformation.

A high degree of variability in the performance of narrow-dimension nickel titanium archwires can be expected being influenced by manufacturer, dimension and degree of tooth displacement. Moreover, susceptibility to deformation appear to vary within batches of identical wires.

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