COMPARISON OF THE MECHANICAL AND SURFACE PROPERTIES OF RETRIEVED AND UNUSED AESTHETIC ORTHODONTIC ARCHWIRES

BY

JENNY FLANAGAN

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School of Dentistry
St. Chads Queensway
Birmingham
B4 6NN
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Abstract

The appearance of orthodontic appliances has been improved by the introduction of tooth coloured brackets. Aesthetic archwires are highly desirable to complement aesthetic brackets in clinical orthodontics. The objective of this study was to characterise the elastic behaviour of aesthetic archwires and determine whether their behaviour was modified following intra-oral use. The load-deflection behaviour of five types of coated and uncoated 0.014” NiTi archwires in their as-supplied condition and following 6 weeks of intra-oral use was characterised using 3 point wire bending tests (n=10 per group). Representative archwires from each group were examined using Scanning Electron Microscopy. The results indicated that the archwires behaviour after 6 weeks of clinical use was more unpredictable than that of unused archwires. In addition the retrieved PTFE coated archwires produced lower unloading forces than unused archwires. The force generated by the uncoated and PTFE coated archwires was comparable. The coated archwires undergo significant delamination of the coating after 6 weeks of clinical use. Within the confines of the limitations of this study the clinical implications are that the PTFE coated archwires moved teeth at comparable rates to their uncoated counterparts. However the coating did degrade and the archwires behaved less predictably after 6 weeks of clinical use.

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Chapter 1

1.1 INTRODUCTION

Orthodontic treatment is usually carried out using fixed appliances that are directly attached to the teeth. The orthodontic brackets, archwires and auxiliary components that make up the fixed appliance mediate tooth movement by controlled application of forces at the tooth-bracket interface. The fixed-appliance components which include the brackets and archwires are routinely manufactured from metals, however there is increasing demand from patients for more cosmetic 'less visible' appliances. As a consequence 'tooth coloured' brackets manufactured from polyurethane, polycabonate and aluminium oxide have been developed, however the aesthetic outcome remains limited due to the visibility of the metal orthodontic archwire. Replacement of the metal orthodontic archwire with a 'tooth coloured' substrate is challenging as the mechanical properties of the archwire itself are fundamental to providing the correct forces to direct orthodontic tooth movement. Accordingly there have been many attempts to camouflage existing metal archwires with tooth coloured coatings to meet the patient's cosmetic demands. The behaviour of these coated archwires in the short term and during usage has not been fully characterised. The objective of this study was to characterise the elastic behaviour of aesthetic archwires and determine whether the behaviour was modified following intra-oral use.

1.2 ORTHODONTIC ARCHWIRES

1.2.1 Levelling and aligning the arches

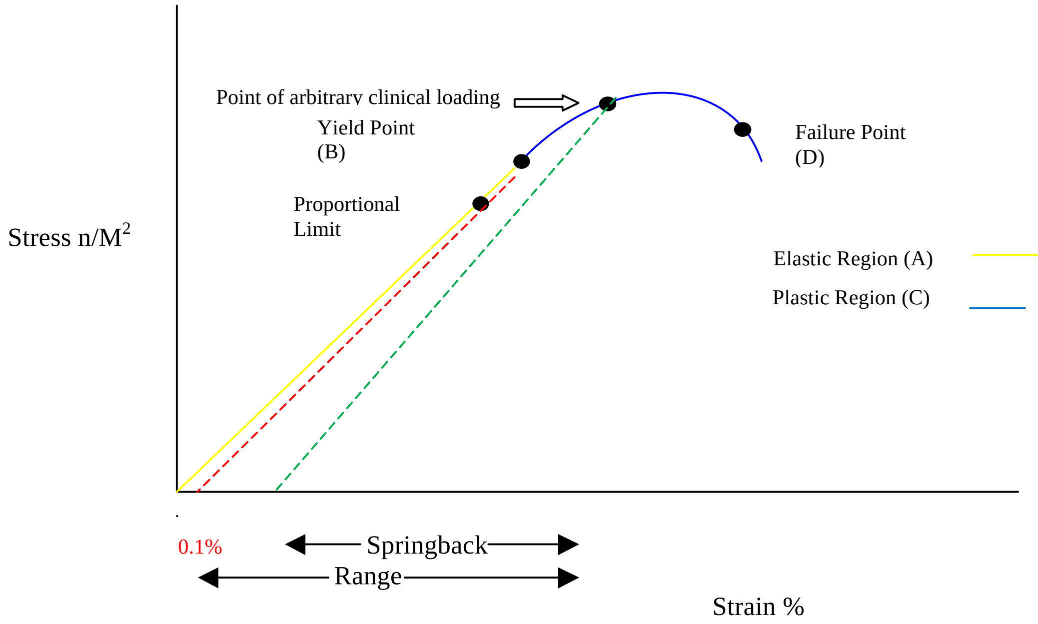
The overall aim of orthodontic treatment is to move teeth into an idealised relationship described by Andrew's six keys (Andrews, 1972). The goals of the first phase of orthodontic treatment are to both align the teeth and to correct vertical discrepancies by levelling out the dental arches. To align the teeth it is necessary to bring malpositioned teeth into the dental arch; to specify and control the antero-posterior position of incisors; the width of the arches posteriorly and the form of the dental arches. Levelling of the dental arch can occur by extrusion of the posterior teeth, by intrusion of the incisors or a combination of both actions (Profitt *et al*, 2007).

1.2.2 Elastic properties of orthodontic archwires

The elastic behaviour of orthodontic archwires is critical to their function and allows the reproducible application of light forces to the tooth to initiate or maintain tooth movement (Kapila and Sachdeva, 1989). The elastic behavior of a material describes its ability to be reversibly deformed and can be identified by measuring the stress-strain relationship generated by externally loading the material (Collings, 1984). Stress can be considered as the internal distribution of the applied load and is defined as force per unit area, measured in Pascal (Pa) where one Pa is equivalent to one Newton per square metre (N/m^2). Strain is the internal distortion of the material as a response to loading and is defined as the magnitude of deformation relative to the material’s original geometry. Strain is considered elastic if it completely reverses when an applied load is removed or plastic where permanent deformation of the material occurs despite the removal of the external load (Kusy, 1997). As strain is a ratio of change in length relative to the original length of a material it is dimensionless and expressed as a percentage.

By studying the stress-strain behaviour of a material it is possible to differentiate the separate regions of elastic and plastic strain (Figure 1). The elastic region is identified as the linear portion of the stress-strain curve and its gradient (stress/strain) allows the calculation of the materials modulus of elasticity or “Young’s modulus” (Pa). The transition from elastic (recoverable) strain and plastic (unrecoverable) strain occurs after the material's yield point. In the context of orthodontic archwires this is particularly important as beyond this point the predictable behaviour of the wire diminishes.

Figure 1.1. is an illustrative plot of the stress-strain relationship of a stainless-steel archwire material under tensile loading demonstrating the elastic region (A), yield point (B), plastic region(C) and failure point (D).



Whilst the strength of a material is often seen as a key material property and frequently reported in the context of orthodontic archwires strength is rarely relevant as catastrophic wire failures rarely occur. In contrast the elastic behavior of the wire and the transition into plastic deformation is of greater interest as they determine the magnitude of force applied to the tooth and therefore a number of features of the stress-strain relationship are commonly reported.

<u>Proportional Limit</u>

The proportional limit is the most conservative measure of the transition from elastic to plastic deformation. It is the highest point where stress and strain still have a linear relationship and this relationship is known as Hooke’s law (Ireland and McDonald, 2003). For the purposes of selecting materials for use as orthodontic archwires it is important to be able to estimate the proportional limit as it identifies the maximum deflection at which the archwire will return fully to its original dimension on removal of the load. It is however extremely difficult to experimentally determine this point precisely.

Yield Strength

The yield strength is a more relevant indicator of the ‘strength’ of a material. It is found as the intersection of the stress-strain curve with a parallel line offset at 0.1% of the elastic strain (Ireland and McDonald, 2003). Typically the true elastic limit lies between the proportional limit and yield strength and both are good estimates of how much force or deflection a wire can withstand clinically before permanent deformation occurs. Once the archwire becomes plastically deformed the load it delivers to the teeth is unpredictable.

Stiffness

‘Stiffness’ is a measure of the resistance offered by an elastic body to deformation (bending, stretching or compression). In the context of orthodontic archwires it is considered as a measure of its resistance to bending and represents the magnitude of force delivered by the archwire at a given deflection within its elastic range (Kusy, 1997). ‘Stiffness’ and ‘springiness’ are terms frequently used within clinical orthodontics and have reciprocal properties where ‘springiness’ = 1/’stiffness’. Each term is proportional to the gradient of the elastic portion of the force-deflection curve. The more horizontal the slope of the force-deflection plot is, the springier the wire is, and the more vertical the slope the stiffer the wire.

Range and Springback

The range is considered as the distance the wire will bend elastically before permanent deformation occurs. If the archwire is deflected beyond this point it will not return to its original shape but can retain clinically useful ‘springback’ unless the failure point is reached (Kusy, 1997). As orthodontic wires are often deformed beyond their elastic limit in clinical practice it is often impossible to determine the exact force required to generate a set deflection. As a consequence springback (the measure of recoverable deformation after the yield point has been exceeded) is an important property in determining clinical efficacy (Kusy, 1997).

Resilience

Resilience is the energy storage capacity of the wire and is the integral of the stress-strain relationship. It is calculated as the area under the stress-strain curve until the proportional limit is reached. Resilience represents the maximum energy per unit volume that can be stored in the elastic region and the amount of energy stored in the wire before it is plastically deformed (Kusy, 1997).

Archwires are presented in different forms and as a generalisable rule the mechanical properties of the archwire are dependent on the material composition, the wire’s cross-section and the wire length:

As the cross-section of the wire is increased

- The range decreases proportionally.
- The springiness decreases as a fourth power function.

As the length of a wire supported at one end is increased:

- The range increases as a square function.
- The springiness increases as a cubic function.

1.2.3 The use of aligning and working orthodontic archwires in fixed appliance treatment

Ideally archwires are designed to move teeth by applying light (0.3 - 1.0 N) continuous forces, although the optimum force magnitude for orthodontic tooth movement has yet to be identified (Ren *et al.*, 2003). It is however accepted that light forces are required to reduce the potential for patient discomfort, tissue hyalinization and undermining resorption (Chan and Darendeliler, 2003). Force application should result in elastic behaviour of the archwire which is required to be maintained over a period of weeks to months (Miura *et al.,* 1986). Different elastic behaviour of the archwire is required according to the treatment stage and the desired tooth movement. As a consequence archwires are fabricated from a range of alloys including stainless steel, cobalt-chromium, nickel-titanium and beta-titanium (Kusy, 1997). Each alloy system and subdivision has different elastic properties and characteristics that may be more appropriate at a particular treatment stage and in contemporary orthodontic practice no single wire is superior for all of the stages of treatment (Kusy, 1997).

It is currently believed that the initial archwires used early in orthodontic treatment for alignment should provide light continuous forces of approximately 50 grams force (0.5 N) to produce the most efficient tooth movement (Ren *et al.,* 2003). In terms of archwire ‘properties’ to enable effective tooth alignment during the initial stages of treatment the archwire requires:

- Low stiffness so that all or most of the teeth in the arch can be ligated.

- Large springback to allow for large deflections of the archwire.
- High stored energy (resilience) so that the archwire returns back to its original shape as the teeth move.
- Biocompatibility to avoid adverse reactions to the archwire material.
- Low surface friction to allow for quick tooth movement.

Of the archwire materials available, nickel-titanium (NiTi) (with a circular cross-section) and stainless steel multistrand or coaxial wires meet most of these material requirements. Typically the alignment wires are used with progressively increasing wire cross-sectional diameters from 0.012 to 0.018 inches depending on the irregularity associated with the dentition. The alignment wires are often followed by a short period (2 - 4 months) of use of a rectangular NiTi archwire to begin torque expression and begin root movement (Cobourne and Dibiase, 2010).

Once initial alignment has been achieved wires of increasing stiffness are then selected to complete the levelling of the dentition, overbite reduction and tooth movement along the archwire. During the later stages of treatment, the process of overbite reduction is completed and if necessary space closure is carried out using sliding mechanics. The archwire used in this stage of treatment generally requires:

- High stiffness to allow for sliding of the teeth along the archwire and space closing mechanics.
- Low stored energy so the wire does not deform under the forces used.
- Biocompatibility avoid adverse reactions to the archwire material.
- Low surface friction to allow for quick tooth movement.

- Good joinability to allow auxillaries such as crimpable hooks to be added on to the archwire.

Rectangular stainless steel wires, known as working archwires, are usually selected at this treatment stage (Cobourne and Dibiase, 2010).

1.2.4 Archwires Materials

1.2.4.1 NiTi as an archwire material for levelling and aligning the arches

1.2.4.1.a History and background

NiTi was the first titanium alloy to be applied as an orthodontic material and its typical composition is 55% nickel and 45% titanium (Andrease and Hilleman, 1971). In the 1960's, the Office of the Navy (USA) was actively studying new types of alloys that exhibited a shape memory effect (SME) (Buehler *et al* 1963). A NiTi alloy showed great promise and was named Nitinol as an acronym for the Nickel-Titanium Naval Ordance Laboratory (Kusy, 1997). The opportunity to use Nitinol in orthodontics was recognised by George Andreasen and through his efforts the first NiTi archwire was marketed to orthodontists under the same Nitinol name (Andreasen and Hilleman, 1971). Today there are three classes of commercially available NiTi wires with different stress-strain relationships: (i) martensitic stable (conventional alloy), (ii) austenitic active (pseudoelastic) and (iii) martensitic active (thermoplastic) (Kusy, 1997).

1.2.4.1.b Properties of nickel titanium alloys

NiTi alloys have two remarkable properties that make them ideal for use as orthodontic archwires, namely shape memory and superelasticity (Andreasen and Morrow, 1997). In common with many alloys NiTi can exist in more than one form or crystal structure and typically for NiTi a martensitic form exists at low temperatures and an austenite form exists at higher temperatures. The key difference between NiTi and other archwire materials is that a transformation from a martensitic to austenitic phase can occur at relatively low temperatures and also can be induced by application of external loading (Miura *et al*., 1986). The two NiTi phases confer different physical and thermal properties to the archwire (Table 1).

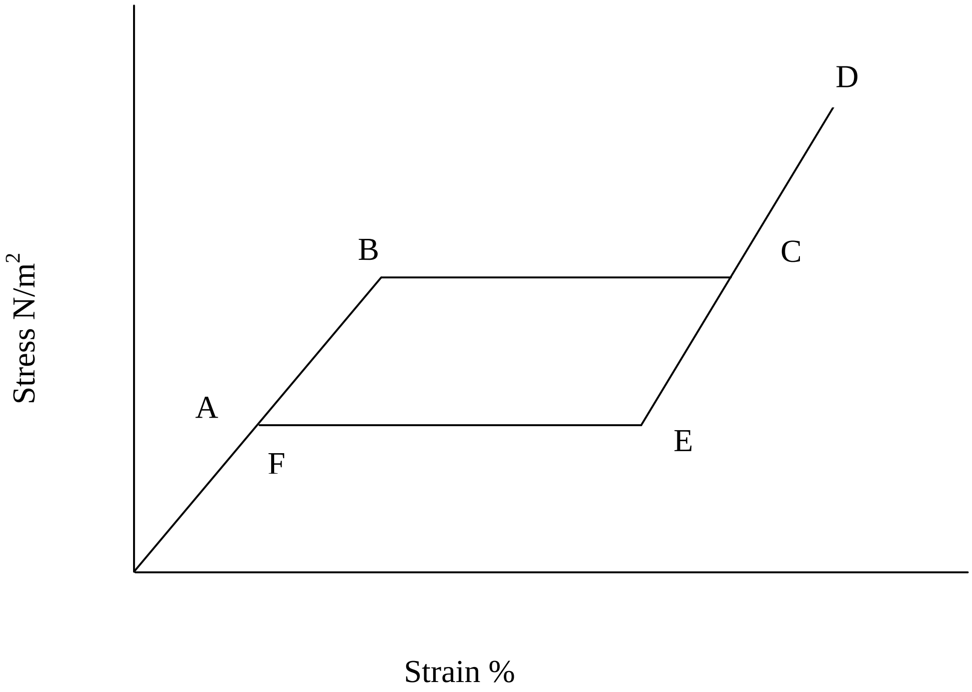
Table 1.1: The physical differences between marsenite and austenite phases

Property	Martensite	Austenite
Yield Strength	1.4 – 1.7 GPa	0.84 GPa
Elastic Modulus	31-35 GPa	84-98 GPa
Thermal Conductivity	0.085 Watt/cm-°C	0.18 Watt/cm-°C
Thermal Expansion	6.6×10^{-6} /°C	11×10^{-6} /°C

Figure 1.2 illustrates an idealised stress-strain curve for NiTi wire. As demonstrated in the stress-strain relationship plot, the archwire begins in an austenitic form and as it is deformed it demonstrates linear elastic behaviour (A-B). As deformation increases the wire undergoes a change in its crystal structure towards the martensitic phase. As the stress remains the same within the wire during this transformation a plateau is seen in the stress-strain relationship (B-C). This is known as a 'stress-induced phase transformation'. Once the alloy has reached a martensitic phase throughout it begins to demonstrate linear elastic behaviour again (C-D). Following a reduction in deformation (such as following tooth movement) (D-E) the alloy transforms back to its austenitic phase and the stress plateaus (E-F). With continued unloading, linear elastic behaviour will once again be demonstrated by the wire (F-A). From a clinical perspective the unloading plateau is especially useful as it means that the wire will apply a light, continuous force to the teeth even though the teeth are moving and the deflection of the wire is reducing significantly. Eventually the stress-induced phase

transformation will have reversed and the austenitic structure will have returned. With continued unloading, linear elastic behaviour will once again be demonstrated by the wire (F-A).

Figure 1.2: Idealised stress-strain curve for nickel-titanium wire showing the plateaus that occur during stress-induced phase transformation.



In addition to stress induced transformations, temperature induced transformation between martensite and austenite phases can be induced by cooling (from austenite to martensite) or heating (from martensite to austenite) the archwire (Kusy, 1997). Intermediate phases are also known to exist including the R phase which exhibits rhombohedral symmetry and a simple hexagonal lattice (Khier *et al.,* 1991). The transformation can therefore be more precisely considered as Austenite to R phase to Martensite on cooling and the reverse if heated (Bradley *et al.,* 1996). However, not all transformations pass through the R phase and the R phase has a lower elastic modulus than austenite (Bradley *et al.*, 1996). Importantly the shape memory and super-elasticity of NiTi are related to phase transitions (between the martensitic and austenitic forms) that occur at a relatively low transition temperature (Profitt *et al.,* 2007) so that these behaviours can be potentially exploited by temperature changes which intercept those encountered in the oral environment.

Early wires had their crystalline structure stabilized or fixed in the martensitic form by cold working and exhibited greater flexibility but no shape memory (Cobourne and Dibiase, 2010). Shape memory refers to the ability of the material to 'remember' its original shape after plastic deformation has occurred and in NiTi this remarkable property occurs whilst the material is in a martensitic phase. The deformation is set while the alloy is maintained at an elevated temperature that is above the martensite-austenite transition temperature. When the alloy is then cooled below the transition temperature it can be plastically deformed but when it is heated again the original shape is restored.

As an orthodontic archwire material Nitinol has a low stiffness and delivers only one-fifth to one-sixth the force per unit of deactivation when compared with an equivalent dimension stainless steel wire (Kusy 1997). In addition the strain that can be applied to the wire before it reaches its yield strength is much greater than for stainless steel. As a consequence functionally when compared with steel, NiTi archwires of equivalent size can move a tooth with a lighter force and for a longer period of time. However a limitation of NiTi is its lack of formability (Ireland and McDonald, 2003).

1.2.4.1.c Superelastic NiTi

Superelastic NiTi is an active NiTi archwire. On loading the austenitic active alloy starts with stiffness that produces three times the force per activation of the conventional martensitic stabilized alloy (Miura *et al.,* 1986). The effect is short-lived and the stress within the wire then plateaus due to a stress-induced phase transformation which occurs from the austenitic to the martensitic form. Upon deactivation (reduction in loading) the martensitic phase gradually transforms back to the austenitic phase and the stress-strain behaviour is typically seen as a rapid drop in stress (associated with unloading) followed by a plateau at low magnitude of stress as the deformation reverses (Miura *et al.,* 1986). When the transformation is complete the stress-strain behaviour of the wire returns to the original behaviour. An orthodontist can exploit this behaviour to align the teeth provided the archwire is activated within the plateau region. The term superelasticity is often used interchangeably in the context of NiTi with the term pseudoelasticity which refers to the second plateau region in which the martensite reversibly transforms to the austenite form and allows low magnitude forces to be maintained (Kusy, 1997).

As stated previously the forces generated in NiTi wires are sensitive to temperature. Filleul and Jordan, (1997) tested four superelastic archwires and reported that three out of the four of the wires showed superelasticity at 22°C, two out the four at 39°C and none of the four at 44°C. The origin of the superelastic plateau was also observed to start at different force levels according to the temperature and lower temperatures were demonstrated to generate a plateau at a lower force (Nakano *et al.,* 1999; Gurgel *et al.,* 2001). Importantly Bolender *et al.,* (2010) suggested that most NiTi archwires do not exhibit the same superelastic behaviour in torsion as traditionally described in flexure (bending) which has clear consequences in the context of orthodontic alignment.

1.2.4.1.d Thermoelastic NiTi

Thermoelastic NiTi wires exhibit a thermally induced shape memory effect. A stress induced phase transformation of the alloy's crystal structure occurs at a temperature range referred to as the temperature transition range (TTR) (Miura *et al.*, 1986). The wire exists in a martensitic phase below the TTR and can therefore be deformed and very easily to be ligated to a severely malpositioned tooth. As the wire heats up a phase transformation occurs to the austenitic phase and the archwire effectively becomes stiffer increasing the load (force) on the tooth and encouraging it to move. The wire has a temperature transition range similar to mouth temperature (35 °C) and hence after ligation and thermal equilibration in the mouth the phase transformation begins. Typically the wire is cooled to facilitate insertion into the brackets of misplaced teeth before undergoing the aforementioned phase transformation and applying force to the teeth through this shape memory effect (Kapila and Sachdeva, 1989).

1.2.4.1.e NiTi Copper Chromium Alloys

The addition of copper to NiTi alloys has been shown to increase strength, reduce hysteresis and allows greater precision in the setting of the austenitic transformation temperature (Gil and Planell, 1995)). The addition of copper also increases the transformation temperature to above that of the oral cavity and therefore requires the addition of 0.2 to 0.5% chromium to reduce the transformation temperature back to a functionally useful range. Originally CuNiTi wires were produced with four different austenitic transformation temperatures (TT) covering both superelastic and thermoelastic archwires: TT Type 1 15°C, Type 2 27°C, Type 3 35°C and Type 4 had a TT of 40°C. A study by Pandis *et al.*, (2009) comparing 0.016" 35° CuNiTi and 0.016" NiTi archwires found no difference in the resolution of mandibular anterior crowding between the two groups.

1.2.4.1.f Clinical effectiveness of NiTi archwires

Numerous studies have attempted to establish the clinical effectiveness of NiTi wires compared with other initial aligning archwires but no studies to date have clearly identified the supposed benefits (O'Brien *et al*, 1990; West *et al*, 1995; Evans *et al*, 1998). It has been postulated that the wires only behave in a superelastic manner when they are subjected to large deflections and this may be why previous studies have failed to establish their superiority in the initial stages of alignment.

1.2.4.2 Beta-titanium archwires

Beta-titanium archwires are also in clinical use a typical composition is 80% titanium, 10% molybdenum, 6% zirconium and 4% tin (Kusy,1997)). Beta-titanium has an advantage over NiTi possessing good formability however the stiffness is roughly one-third that of stainless steel and half that of NiTi (Burstone and Goldberg, 1980). In addition beta-titanium archwires are also associated with higher friction and undergo smaller levels of permanent deformation. They are often used in the final stages of treatment, when finishing bends may be required to detail individual tooth position and achieve settling of the occlusion (Profitt *et al.,* 2007).

1.3 AESTHETIC ORTHODONTIC ARCHWIRES

1.3.1 The demand for aesthetic archwires

The demand for aesthetic orthodontic appliances is increasing and the aesthetics of orthodontic appliances has improved significantly with the use of transparent brackets fabricated from ceramic or composite materials (Russell, 2005). Other approaches for aesthetic orthodontic treatment include aligners and lingual appliances (Lagravere and Flores-Mir, 2005; Poon and Taverne, 1998; LY, Kula K, 2006). A recent study by Feu *et al.,* (2012) found that clear aligners were considered to be the best aesthetic option by adults, followed by the combination of sapphire (aluminium oxide) brackets and aesthetic archwires. The demand for aesthetic appliances is driven by the increased number of adults seeking orthodontic treatment. Aesthetic archwires are highly desirable to complement aesthetic brackets in clinical orthodontics.

1.3.2 History

A number of alternatives to metal have been explored to create an aesthetic archwire that would allow efficient orthodontic treatment with the appliance visible labially (Russell, 2005). Archwires and in particular NiTi have been coated in tooth coloured polymers or inorganic materials. Deficiencies with coated archwires include the fact that the coating is frequently reported to peel or wear and it can be difficult to bend the wire (Neumann *et al.,* 2002). The extent of the coating is limited by the small cross sections needed for orthodontic wires. In addition since the outer surfaces of the archwire are most distant from the neutral axis they are biomechanically the most important and should therefore have optimal properties and hence placement of a coating may affect archwire efficacy.

Polymeric composite wires have been in development since the early 1990s. Their advantage over coated wires is their transparency and their capability to vary the stiffness of the wire without changing its cross-sectional profile (Burstone, 1981). There is a debatable concern regarding metallic alloys and allergic reactions to nickel (Jones *et al.*, 1986). Some authors have found no effect on patients, others described a sensitization of the patients by the use of nickel containing materials and several authors reported cases of allergic reactions to nickel containing orthodontic devices and advise the use of nickel free devices for the treatment of patients already sensitized to nickel (Vreeburg *et al.,* 1990; Kusy *et al.,* 1998). This problem could be avoided with composite materials (Jones *et al.*, 1986). The problems associated with polymeric archwires are that although they have many of the properties necessary for an orthodontic archwire they have insufficient rigidity and strength.

1.3.3 Development of Aesthetic Archwires

1.3.3a Metal-free archwires

There have been considerable efforts to develop totally metal-free alternatives to currently available metallic archwires. The first aesthetic non metallic orthodontic wire, Optifelx, contained a silica dioxide core that provided the force to move the teeth, a silicone resin middle layer that protected the core from moisture and a stain-resistant nylon outer layer which serves the dual purpose of preventing damage to the archwire and further increasing the strength of the archwire (Talass, 1992). Advantages of this wire include very good aesthetics and it's resistance to stains however it is brittle and sharp bends may result in fracture of the glass (SiO_2) core. It has been characterised as possessing poor springback and thus its clinical efficacy and ability to be used in a variety of cases has been questioned (Lim *et al.,* 1994).

Subsequently an archwire with S2 glass fibres embedded in a polymeric matrix formed from bisphenol A-diglycidylether methacrylate and triethylene glycol dimethacrylate was developed by Fallis and Kusy (Fallis and Kusy, 2000). Other research groups have developed different fibre reinforced polymer wires however although the appearance of the wires is excellent clinical uptake has been poor because of their brittle nature (Zufall and Kusy, 2000). They are also very strong and the stiffness can vary from that of the most flaccid multi-stranded archwire to nearly that approaching the properties of a beta-titanium archwire. Mechanical tests show that these archwires are elastic until failure occurs and resilience and springback are comparable with NiTi (Fallis and Kusy, 2000). When the wire fails, it loses its stiffness but remains intact. As composites are displacing metallic alloys as structural components in the aerospace industry, the expectation is that the attractive properties and characteristics of these materials will capture a significant share of the marketplace in the future.

Recently Burstone described a polymer archwire that had polyphenylene polymer extruded into it (Burstone *et al.,* 2011). Polyphenylene enhances rigidity, strength and hardness as well as increasing resistance to stress relaxation. The resultant wire delivered forces generally similar to typical beta-titanium and NiTi wires of somewhat smaller cross sections. The magnitude of force the wire produced was similar to that of an aligning or levelling wire. This wire could serve as a potential aesthetic archwire in the future but further research is needed (Burstone *et al.,* 2011).

1.3.3b Coated archwires

The inability to match the mechanical properties of metallic archwires using non-metal alternatives has resulted in the development of coated archwires which provide a compromise allowing some cosmetic improvement whilst aiming to maintain the desirable mechanical properties of the metallic archwire.

Teflon coated archwires

An obvious materials development to improve the aesthetics of metallic archwires whilst retaining the functionality includes the application of cosmetic surface coatings. The most common commercially available coated archwires use polytetrafluroethylene (PTFE) –a material which has numerous applications in medicine and dentistry (Chiapasco, 1999). PTFE (Teflon) has also been advocated as a material to be used to improve the anti-cariogenic properties of composite resins (Gyo *et al.,* 2008), coatings of metallic stents for palliation of malignant biliary disease (Hatzidakis *et al.,* 2007), artificial muscles and as a conduit for guided nerve regeneration (He *et al.*, 2003). PTFE is a polymer with a completely fluoridated chain which confers both its physical and chemical characteristics. The physical properties of PTFE are shown in Table 2 (Farranoto *et al*, 2012).

Table 1.2 Physical characteristics of Teflon (PTFE).

Properties	Value
Molecular Weight	5 x 10 to 5 x 10^6
Density	2170kg/m^3
Softening temperatures	615 K
Fusion temperatures	6000 K
Modulus of elasticity	0.41 – 0.55GPa
Load at failure	14 – 28 MPa
Èlongation at break	100 - 400%

PTFE coatings are applied by an atomic process, using compressed air to deposit a layer of about 20-25 μm thickness on the wire which then undergoes a heating process in a chamber furnace (Husmann *et al.*, 2002). The sintering fabrication results in two common coating forms: classical PTFE which is not microporous (Teflon) and expanded PTFE which is microporous and characterized by oriented microfibrils kept together by solid junctions. PTFE coatings impart a tooth coloured hue which is marketed as being similar to that of natural teeth.

Rhodium coated archwires

Rhodium is a hard, silvery-white transition metal that is a member of the platinum group. 80% of its use is as one of the catalysts in the three-way catalytic converters used in automobiles (Loferski, 2013). It is usually alloyed with platinum or palladium and applied in high-temperature and corrosion-resistant coatings. Rhodium coated archwires became available for commercial use in 2008. These wires have low reflectivity which is promoted as conferring reduced visibility and improved aesthetics.

1.3.4 Contemporary coated aesthetic archwires

PTFE coated archwires are currently available in thicknesses that range from 0.5 mm to 0.02 mm and come fully coated or with a layer of coating on the labial surface only. They are also available with a layer of coating just on the anterior segment of the archwire. Sentalloy and Bioforce (Dentsply GAC, Canada) high aesthetic rhodium coated superelastic archwires are available with a 0.127 mm rhodium coating.

In 2011 a new aesthetic archwire called Woowa was introduced with a currently compositionally undeclared polymer coating that is 0.000127 mm thick and tooth-coloured (Iijima *et al.,* 2012). It is reported as having a double layered coating structure, the inner layer is silver and platinum and the outer layer is polymeric. The concept is that the outer layer imparts durability and wear resistance and the inner layer gives the archwires its tooth-coloured appearance. It is available as a coating on both NiTi and stainless steel archwires.

1.4 EVIDENCE FOR THE PERFORMANCE OF AESTHETIC ORTHODONTIC ARCHWIRES

1.4.1 Mechanical Properties

Loading and unloading forces produced by PTFE coated archwires have been reported to be lower than uncoated archwires (Bradley et al, 2013; Alavi and Hosseini, 2012; Iijima *et al*, 2012; Kaphoor *et al*, 2011; Elayyan *et al*, 2010; Elayyan *et al,* 2008). This may be due to reduced diameter of the NiTi wires occupying the inner core of the aesthetic wires. A more recent study did however find that four out of the eleven coated archwires tested, exhibited similar unloading force values when compared with their uncoated counterparts (Washington *et al.,* 2014). Although lower unloading forces may be desirable, they may not be effective if it is below the optimum range for orthodontic tooth movement. Clinicians may need larger wires to get the equivalent force value which increases friction. Friction between the wire, bracket and ligature may also be important in determining the amount of force delivered by the wire and as the deflection increases so the angle of emergence of the wire from the bracket becomes more acute. This increased friction reduces unloading and increases loading forces and is likely to lead to more damage to the coating.

Elayyan *et al.,* (2008) found that the unloading forces generated by retrieved epoxy resin-coated archwires with conventional ligation that have been used in vivo for 4 – 6 weeks are lower than as-received coated archwires. In contrast a more recent study by Bradley *et al* (2013) found that retrieved coated archwires produced higher unloading forces when compared to as-received archwires. They attributed this to the fact that as the coating was lost the wires behaved more like their uncoated couterparts. When self-ligating brackets are used, the force value for retrieved archwires is the same as the as-received coated archwires. This may occur because the self-ligating brackets have reduced friction and are therefore not as affected by the delamination and increased friction of the retrieved archwires.

1.4.2 Surface Topography

Coated archwires have been shown to have greater surface roughness than uncoated archwires (Doshi and Wasundhara, 2011; Zufall and Kusy, 2000). Mixed results have been reported about the relationship between surface roughness and friction. Some studies demonstrated that there was a positive correlation between surface roughness and frictional resistance, however other studies reported there is little correlation between surface roughness and friction

(Saunders and Kusy, 1994; Doshi and Wasundhara, 2011; Farronato *et al.*, 2012). The coating layers might influence the frictional characteristics of coated wires but further research is needed to prove this hypothesis.

Retrieved NiTi archwires have been found to be covered with islands of amorphous precipitants and accumulated microcrystalline particles. The wires have a proteinaceous biofilm whose organic constituents are mainly amide, alcohol and carbonate. Elemental species precipitate on the material surface and include NA, K, Cl, Ca and P, forming NaCl, KCL and Ca-P precipitates. Porosity and size pore of the wire surface also increases (Grimsdottir and Hensten-Petersen, 1997).

Coated archwires demonstrate rougher surfaces after use in vivo compared with as received archwires (Rongo *et al.*, 2013; Wichelhaus *et al.*, 2005; Neumann *et al.*, 2002) which may be due to the coarse influence of tooth brushing and the interplay of the archwire coating and bracket edges. Imprints of brackets and areas of degradation have been found in areas related to the positions of the brackets, and these surface defects may hinder the archwire sliding through the bracket (Neumann *et al.*, 2002).

1.4.3 Durability

It is noted that coatings on aesthetic archwires had a tendency to split and peel off during use (Proffit, 2000). Elayyan *et al.*, (2008) found that retrieved PTFE coated archwires suffered from inconsistent amounts of deterioration after a mean of 33 days in vivo. Many of the retrieved specimens were characterized by delamination of the coating over large areas. Although the coating remained intact in some areas, it showed a rougher, discoloured and deteriorated surface when compared with unused archwires. In total 75% of the PTFE coating remained on the archwires which is likely to affect the aesthetic value of these aesthetic archwires (Elayyan *et al.*, 2008). Similar results have been reported in more recent studies with common objectives (Bradley *et al*, 2013; Da Silva *et al*, 2013; Alavi and Hosseini, 2012).

Aesthetic archwires can be discoloured by external sources such as food dyes and coloured mouth rinses. The extent of discolouration depends on the type of coating material and its surface roughness, the level of oral hygiene and water absorption (Faltermeier *et al.,* 2008). Da Silva *et al.,* (2013) assessed the colour stability of six aesthetic archwires at different time periods using a staining coffee solution. The investigators found that all the aesthetic archwires showed clinically noticeable colour change after 21 days in the staining solution. The fibre-reinforced metal-free Optis archwire had the most colour alteration, followed by the coated NiTi and stainless steel archwires (Da Silva *et al.,* 2013).

1.4.4 Fatigue

The life expectancy of NiTi archwire applications may reach time frames in the order of 1 year. Mechanical fatigue of the archwire is therefore a relevant phenomenon and the effects of cyclic loading; fluctuations in the intraoral environment including pH; biofilm aggregation is all additionally implicated in the aging pattern and typical fracture characteristics of these wires (Bourauel *et al.,* 2008). It has been demonstrated that retrieved archwires fracture at a lower number of loading cycles when compared with their unused matches. Deterioration in the mechanical properties of NiTi archwires has been demonstrated to be more susceptible to fatigue when compared with stainless steel and beta-titanium archwires. The size of the wire also plays an important role in determining the fracture, with larger cross-sections showing reduced fatigue failure properties. Rectangular cross sections possess an increased chance of failure compared with round archwires (Bourauel *et al.,* 2008). It may be necessary to monitor patients to recognize orthodontic wire failures despite the commonly deployed practice of increased time between appointments in patients treated using NiTi archwires.

1.5 ANALYSIS OF RETRIEVED ARCHWIRES – COLLECTION AND MODIFICATION OF PROPERTIES

Retrieval analysis has been used for a number of years in the orthopaedic application of biomaterials. It was first described by Rostoker *et al.,* in 1978 who examined polyethylene components of hip and knee joint prostheses removed from patients. Due to increasing interest and the high number of published studies concerning retrieved orthopaedic materials, standards for the retrieval analysis of orthopaedic materials have been developed. (International Standards Organization/Draft International Standards, 1996; American Society for the Testing of Materials, 1997) however to date no ISO or ASTM standardisation exists for similar objectives in dentistry.

Failure analysis is however gaining popularity in dental materials science due to the significant information resulting from examining the performance of the material in the environment in which it functions (Eliades *et al.,* 2000; Bourauel *et al.,* 2008; Daems *et al.,* 2009; Zegan *et al.,* 2012; Bradley *et al.,* 2014)). It is useful in establishing the effects of intra-oral aging on the behaviour of orthodontic archwires. Intra-oral use may affect the properties of archwires such as superelasticity, fracture resistance, force delivery and friction. This method has also gathered previously undescibed information, such as the assault of specific microbes on orthodontic materials (Matasa, 1995). One disadvantage of these retrieval studies is the absence of sequential description of the changes that were induced and the failure to gather quantitative data.

1.6 METHODS OF ASSESSING THE PROPERTIES OF ORTHODONTIC ARCHWIRES

There are many tests available to assess the mechanical and physical properties of archwires. Although these tests do not necessarily reflect the clinical situation to which wires are usually subjected, they provide a basis for comparison of these properties.

1.6.1 Load deflection properties

Springback can be referred to as maximum elastic deflection, maximum flexibility, range of activation, range of deflection or the working range. It is related to the ratio of yield strength to the modulus of elasticity of the material. Higher springback values provide the ability to apply large activations with a resultant increase in working time of the appliance. This implies that fewer archwire changes will be needed. The load deflection rate is the force magnitude delivered by an appliance and is proportional to the modulus of elasticity. Low stiffness or load deflection rates allow lower forces to be applied, a more constant force over time and the appliance experiences deactivation and greater ease and accuracy in applying a given force (Kapila and Sachderva, 1989). The load deflection properties are measured using three-point wire bending tests that were first described by Miura *et al.* in 1986. The test was designed to demonstrate the difference between the first nitinol wire and the superelastic nickel-titanium archwire. This is thought to be the most significant parameter when determining the biologic nature of orthodontic tooth movement (Krishnan and Kumar, 2004). It provides information on the behaviour of the wires when exposed to both horizontal and vertical deflections (Kapila and Sachderva, 1989). This test offers a high degree of reproducibility which enables comparison between studies (Wilkinson *et al.*, 2002). Other advantages include its close simulation to clinical application and the capacity to distinguish wires with superelastic properties. The orthodontic wire is deflected and the generated load is measured. The load-deflection curve generated is analysed to detect the mechanical properties of the archwires. A fixture between two supports at a pre-determined distance is used. The wire specimen is secured on orthodontic brackets fixed on the poles using elastomeric ligatures and the testing is done using a Universal Testing Machine (UTM). A loading platern is attached to the cross –head of the UTM and the wire is centrally deflected at a fixed deflection rate (most commonly 1mm/min). The loading and unloading values are then recorded at specific deflections. The loading curve represents the force needed to engage the wire in the bracket, whereas the unloading curve represents the forces delivered to teeth.

1.6.2 Surface Topography

The surface topography of orthodontic archwires can be studied with high resolution, microscopy. Scanning Electron Microscopy (SEM) is a type of electron microscope that images a sample by scanning it with a beam of electrons in a raster scan pattern. The electrons interact with the atoms that make up the sample producing signals that contain information about the sample's surface topography, composition and other properties such as electrical conductivity. The images are typically qualitatively assessed although image analysis protocols may enable quantifiable data to be generated.

1.6.3 Surface Roughness

A surface profilometer is a measuring instrument which is used to quantify the features of a surface topology. A diamond stylus (or non contact guage such as a laser of white light interferometer) is positioned over the sample and moved laterally for a specified distance (with a specified contact force for contact methods). Profilometry methods can measure small surface variations in the vertical access as a function of position, ranging from nanometers to millimetres depending on the specific guage resolution. Common quantification parameters of surface roughness include the Ra-value which is a mesure of the “roughness” of the surface and is defined as “the arithmetic mean deviation of the roughness profile from the mean line”.

1.6.4 Fatigue

The fracture resistance of orthodontic archwires can be measured by subjecting the archwire specimen to cyclic mechanical loading in a set-up that simulates intraoral loading at deflections of predetermined amounts. The test is performed at 37°C in doubly distilled, sterile water. The specimen of wire is loaded in a three point bending mode and is fixed at one end only. The other end can move freely and no additional tensile stress to bending deformation is applied. The loading frequency is usually set to 1 or 2Hz and cyclic loading is applied either until fatigue fracture or until a maximum number of loading cycles of 2×10^6 is reached (Bourauel *et al.,* 2008).

1.6.5 Durability of the coating

Ellayan described a method of measuring the amount of coating remaining on the archwire after use using photography. Digital photographs of each side of the wire are taken with the camera fixed on a tripod and oriented at 90° to the surface on which the archwire is placed. The distance from the camera to the wire remains constant and a ruler is placed adjacent to the archwire for calibration purposes. Images are then recorded of the gingival and occlusal aspects of the archwire and transferred to a computer where the Image J program is used for analysis. This programme is first calibrated using the ruler in the photograph and the overall length of the wire and the length of the remaining coated segments is measured. These measurements are exported to Excel and the percentage of the remaining coating is calculated by dividing the sum of the length of the remaining coated segments over the overall length of the wire. The average percentage of the two sides of each wire is then calculated (Elayyan *et al.*, 2008).

1.7 AIMS AND OBJECTIVES

The aim of this study is to assess the mechanical and surface properties of coated archwires before and after clinical use. The specific objectives of this study are to:

1. Determine the force-displacement relationship of unused 0.014" coated nickel titanium archwires through 3 point wire bending tests
2. Determine the force-displacement relationship of 0.014" coated nickel titanium archwires after 6 weeks of clinical use through 3 point wire bending tests
3. Assess the surface of unused coated 0.014" nickel titanium archwires
4. Assess the surface of coated 0.014" nickel titanium archwires after 6 weeks of clinical use

1.8 NULL HYPOTHESIS

Two null hypotheses are considered in this study:

1. There is no difference in the load deflection properties of coated and uncoated superelastic 0.014" NiTi archwires.
2. There is no difference in the load deflection properties of unused and retieved 0.014" NiTi archwires of the same size

Chapter 2 - Materials and Methods

Acknowledgment

This study, registered with the University of Birmingham, was based at the Orthodontic Departments of the Birmingham Dental Hospital and Worcester Royal Hospital. The archwires were sourced by the author and laboratory based studies were carried out in the Biomaterials Unit, University of Birmingham School of Dentistry and the Materials Science Unit, Dublin Dental University Hospital, Trinity College Dublin.

2.1 ETHICAL APPROVAL

Clarification on the requirement for necessity for ethical review was sought from National Research Ethics Service prior to the commencement of the study. A study protocol and self evaluation of the ethical implications according to the NRES “Does my project require review by a Research Ethics Committee” algorithm was provided. The evaluation by NRES confirmed no requirement for ethical review and further correspondence with NRES is included in Appendix1.

2.2 ARCHWIRE MATERIALS

Five different NiTi archwire systems were identified for inclusion in the study. 0.014” NiTi archwires were selected because this dimension of archwire is commonly used as the first archwire after the patient has been bonded up with fixed appliances. As the archwires were to be fitted by several members of staff in two different units it was more convenient to use a 0.014” NiTi archwire as the majority of clinicians at the two sites use this wire dimension as their initial aligning archwire.

The five selected archwires systems were:

Uncoated Archwires

1. 3M Unitek Nitinol Superelastic Archwire – this archwire is superelastic and uncoated. It is manufactured by 3M Unitek Orthodontic Products, Monrovia, CA, USA

   Batch Number: 4296-911 / 4296-814.

2. Sentalloy Archwire – this archwire is pseudoelastic and thermally activated. It is manufactured by Dentsply GAC International, Bohemia, NY, USA and supplied by TOC (The Orthodontic Company), Bristol, UK.

   Batch Number: 02-517-112.

Coated Archwires

3. Euroline Micro-Coated Archwire – this archwire is coated labially with a 0.005” PTFE layer. It is manufactured and supplied by DB Orthodontics Ltd, Silsden, UK.

   Batch Number: 102372/102372.

4. Orthocare Tooth Coloured Archwire – this archwire is fully coated with a 0.002” PTFE coating. It is manufactured and supplied by Orthocare, Ortho-Care (UK) Saltaire, UK.

   Batch Number: 5323-940/5232-5890.

5. High Aesthetic Sentalloy Archwire - It is a pseudoelastic nickel titanium archwire with thermally activated shape memory. It has a 0.002” rhodium coating that reduces its reflectivity and thus is not a tooth coloured archwire. It is manufactured by Dentsply GAC International and supplied by The Orthodontic Company.

   Batch Number: 02-711-112/02-711-143.

Patients were fitted with 3M APC PLUS adhesive coated stainless steel brackets (3M ESPE, St Paul, MN, USA) with 18% chromium and 8% nickel content. The dimensions of the brackets were 0.22” x 0.28” and each bracket possessed a formed mesh foil base to aid retention. The archwires were secured to the brackets with DB Orthodontic elastomeric modules which are composed polyurethane.

2.3 PATIENT SELECTION

The patients included in this study were undergoing fixed appliance treatment at the Birmingham Dental Hospital and the Worcester Royal Hospital. No inclusion or exclusion criteria were provided however all patients had IOTN assessment scores of at least 4 consistent with eligibility for orthodontic care within the two hospital settings. Archwires were fitted in patients following their initial bond-up with fixed appliances in either the maxillary or mandibular arches. Orthodontic care was provided by speciality registrars or consultant orthodontists within the two departments. Archwires were allocated in sequential blocks by wire type (n=10) to the patients as they presented to the orthodontic care providers.

2.4 PLACEMENT AND RETRIEVAL OF ARCHWIRES

All patients were treated with straight wire appliances with MBT bracket prescriptions. The slot dimension of the brackets used was 0.022” x 0.028”. The teeth were prepared with 3M Unitek Transbond self etching primer. The bond was agitated on each tooth surface for 3-5 seconds. It was then lightly blown with non-compressed air away from the gingiva. The brackets were all precoated with composite and positioned on the facial axis of the clinical crown of the teeth and light cured for 20 seconds. The archwires were ligated into place with DB Orthodontic elastomeric modules. No stainless steel ligatures were used and the elastomeric modules were used in a standard configuration.

The archwires were retrieved after six weeks in situ. This time frame was chosen as it is the average amount of time between the first two appointments following placement of fixed appliances. As it was not possible to retrieve all of the archwires exactly six weeks later a tolerance of ± 3 days was given so that all archwires were retrieved between 39 and 45 days in-vivo. The archwires were wrapped in damp gauze sealed in a plastic bag prior to storage at 4°C. The type of archwire, date of insertion and date of retrieval were recorded.

2.5 DETERMINATION OF FORCE-DISPLACEMENT RELATIONSHIPS OF UNUSED AND RETRIEVED WIRES

2.5.1 Wire bending tests

Ninety 0.014" NiTi archwires were subjected to the 3 point wire bending tests. A Sentalloy uncoated archwire was compared with the Rhodium coated high aesthetic Sentalloy archwires in an unretrieved state to acts as a control to study the effect of rhodium coating on the archwire properties.

Archwire type	Retrieved	Unretrieved
3M Unitek uncoated archwires	✓	✓
Sentalloy uncoated archwires		✓
Rhodium coated high aesthetic Sentalloy archwires	✓	✓
Euroline microcoated archwires	✓	✓
Orthocare tooth coloured archwires	✓	✓

Three point wire bending tests were carried out using a Universal Testing Machine (Instron 5544, Instron Ltd, Buckingham) with a calibrated 100 N load cell. A jig was constructed by cementation of precisely aligned central incisor and canine brackets to a levelled aluminium baseplate. The surface of the baseplate was sandblasted with 50µm Al_2O_3 to create a surface to maximise adhesion to the resin composite cement. A 3M APC PLUS adhesive coated stainless steel bracket upper right central incisor and canine bracket of 0.022 x 0.028 inches with an MBT prescription was bonded onto the baseplate using 3M Unitek Transbond self etching primer (3M ESPE, St Paul, MN, USA). The interbracket distance was 14 mm which is equivalent to the average anatomical distance between the central incisor and canine brackets. The jig was secured to the base of the UTM.

Each archwire was sectioned at the midpoint into two pieces. The right archwire specimen was attached to the brackets and ligated using 3M polyurethane elastomeric modules (3M ESPE, St Paul, MN, USA) in a standard configuration to study displacements up to a maximum of 1mm whereas the left archwire specimen was similarly ligated and loaded to a maximum 2 mm displacement. The ligated archwires were centrally loaded using a cylindrical indenter at a crosshead speed of 1 $mm.min^{-1}$. A displacement limit of either 1 mm or 2 mm was set and following loading to this limit unloading at 1 $mm.min^{-1}$ was immediately performed. Tests were conducted at 23 ± 1°C. Loading and unloading forces were recorded with data capture every 15 ms. In total 180 wire specimens from the 9 Groups were tested.

2.5.2 Scanning Electron Microscopy

Representative archwires from each group were examined using Scanning Electron Microscopy (SEM). The archwires were cut into 10 mm specimens and attached to adhesive carbon tabs on aluminium studs. The specimens were then gold sputtered using a deposition current of 20 mA and a deposition time of two minutes using in an Emitech K550x sputter coater (Quorum Technologies Limited, East Grinstead, UK). Images were captured using a Zeiss scanning electron microscope was used (Model EVOMA10-159, Series 51-1385-026, Zeiss GmbH, Jena, Germany) at various magnifications (x100, x1000 and x1500) using an accelerating voltage of 5 kV and a working distance of 7.5 mm for all magnifications.

2.5.3 Surface profilometry

Surface roughness (Ra-value [μm]) of the unused and used archwires was determined using non-contact profilometry (Talysurf CLI 2000, Taylor-Hobson Precision, Leicester, UK). A chromatic length aberration gauge (300 μm range in the z-axis; scanning speed of 500 μm/s) was used to measure line tracts down the long axis of the labial surface of the archwires. A 0.25 mm cut-off Gaussian filter according to ISO 4287 was employed to calculate Ra. One line track per sample was recorded and the mean Ra-value calculated.

2.6 STATISTICS

Comparions between wire types for parameters obtained from load-deflection curves (peak load, load at specified values) were made using a one-way ANOVA and post-hoc Tukey tests (α =0.05). Comparisons for parameters obtained from load-deflection curves (peak load, load at specified values) between individual wire types in their unused and used states were made using unpaired t-tests (α =0.05). Mean Ra-data was compared using a two-way ANOVA where the factors were (wire type and exposure status [used, unused]) (α =0.05). SEM images were qualitatively assessed at standardised magnifications.

Chapter 3 – Results

3.1 LOAD DEFLECTION BEHAVIOUR OF UNCOATED AND COATED 0.014’ NITI ARCHWIRES

The load-deflection behaviour of five types of 0.014” NiTi archwires in their as-supplied unused condition was characterised using 3 point wire bending tests (n=10 per group). Two uncoated archwire types were selected as uncoated controls -3M Unitek uncoated archwires were the uncoated control for Euroline microcoated archwires and Orthocare tooth coloured archwires; Sentalloy uncoated archwires were the control for Rhodium coated high aesthetic Sentalloy archwires.

3.1.1 Load-deflection behaviour of uncoated archwires

Figure 3.1 the mean load and the associated standard deviations (N) required to deflect the 3M uncoated archwires centrally to maximum deflection of 1 mm (Figure 3.1a) or 2 mm (Figure 3.1b) at a constant loading rate of 1mm/min. Once the peak deflection had been reached the unloading was undertaken at a constant rate of 1mm/min.

Figure 3.1a

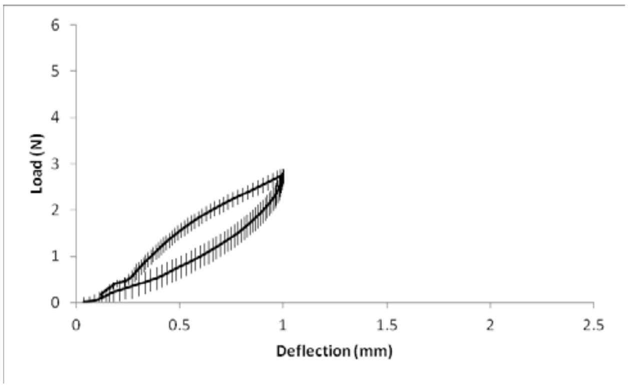
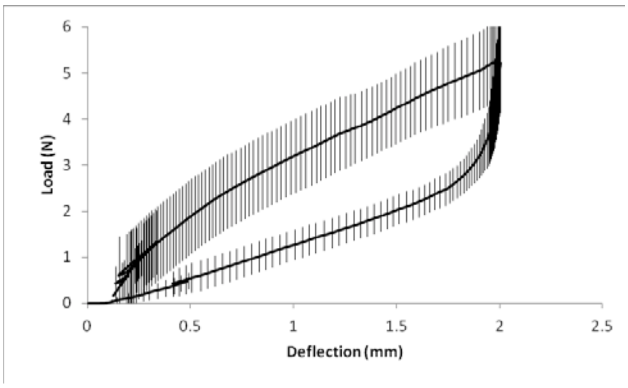


Figure 3.1b



The loading and unloading gradients were observed to be steep to achieve a 1 mm central deflection indicating that the force generated when ligating the wire into the bracket increases rapidly with increased deflection of the archwire and that the force dissipates rapidly when the wire becomes less deflected following tooth movement. For the unused 3M uncoated archwires deflected to 2 mm the initial load deflection gradient was similar to the 1 mm deflection study measured separately and a relatively steep gradient was maintained until the maximum deflection was reached. Large standard deviations were observed as seen in Figure 3.1b illustrating considerable variability in the measurements. In addition the archwire deflection did not return to zero following complete removal of the external load which may be due to the archwire becoming distorted or because of some displacement of the wire at the ligated bracket sites.

Figure 3.2 the mean load and the associated standard deviations (N) required to deflect the uncoated Sentalloy archwires centrally to maximum deflection of 1 mm (Figure 3.2a) or 2 mm (Figure 3.2b) at a constant loading rate of 1mm/min. Once the peak deflection had been reached the unloading was undertaken at a constant rate of 1mm/min.

Figure 3.2a

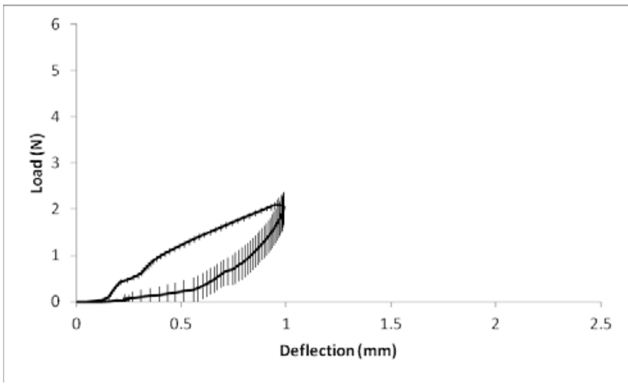
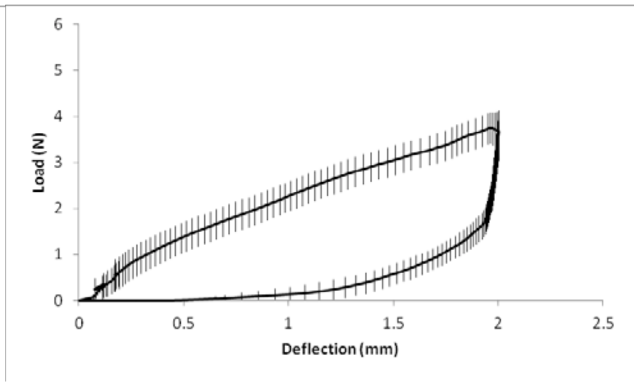


Figure 3.2b



The load deflection behaviour of the Sentalloy uncoated archwires differed from the 3M uncoated archwires with a clear plateau region observed in the unloading profile of wires deformed to both 1 mm (Figure 3.2a) and 2 mm (Figure 3.2b) peak deflections. The plateau regions were however associated with very low force application (<0.5 N). For wires deflected to 1 mm (Figure 3.2a) large standard deviations on the unloading curve were observed and a zero force was reached before the deflection returned to 0 mm. In addition fluctuations are evident on the loading curve and this was not isolated to a single sample. In general there was greater consistency in the measurements taken for the Sentalloy uncoated archwires when compared with the 3M uncoated archwires.

3.1.2 Load-deflection behaviour aesthetic coated archwires

Figure 3.3 the mean load and the associated standard deviations (N) required to deflect Euroline labially coated archwire centrally to maximum deflection of 1 mm (Figure 3.3a) or 2 mm (Figure 3.3b) at a constant loading rate of 1mm/min. Once the peak deflection had been reached the unloading was undertaken at a constant rate of 1mm/min.

Figure 3.3a

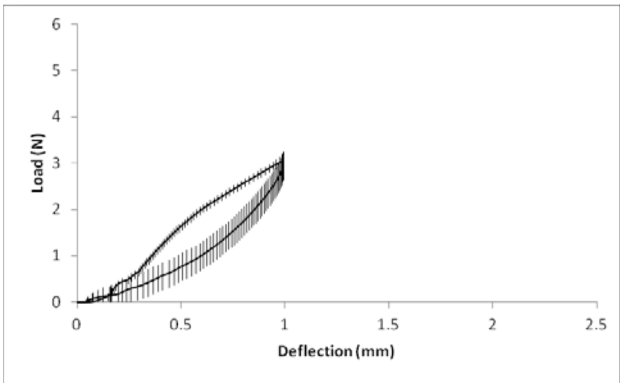
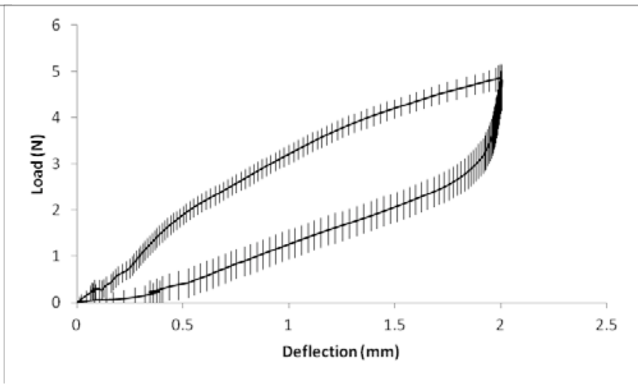


Figure 3.3b



The load-deflection plots of the Euroline labially coated archwires exhibit a similar pattern to the 3M uncoated archwires at both 1 and 2 mm peak deflections. Steep loading and unloading gradients were observed in Figure 3.3a and very little in the way of a plateau region was demonstrated. Large standard deviations were observed in the unloading portion of the load deflection plot. For the Euroline labially coated archwires deflected to a peak of 2 mm (Figure 3.3b) the gradient of the loading curve was observed to decrease with increasing deflection.

Following unloading a rapid reduction in force was observed consistent with the pattern observed in Figure 3.1b. Early during deflection (< 0.25 mm) inconsistencies in the recorded load were observed which could be attributed to movement at the bracket ligature interface.

Figure 3.4 the mean load and the associated standard deviations (N) required to deflect Orthocare fully coated archwire archwire centrally to maximum deflection of 1 mm (Figure 3.4a) or 2 mm (Figure 3.4b) at a constant loading rate of 1mm/min. Once the peak deflection had been reached the unloading was undertaken at a constant rate of 1mm/min.

Figure 3.4a

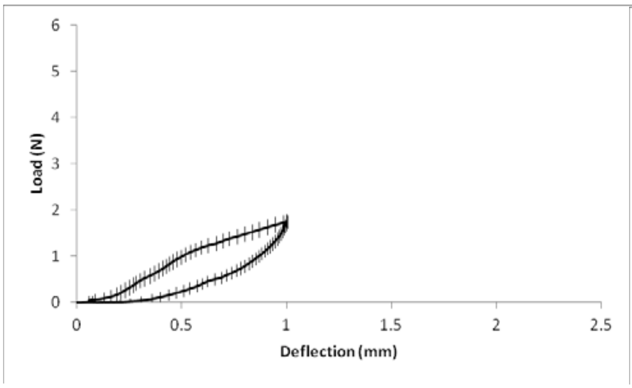
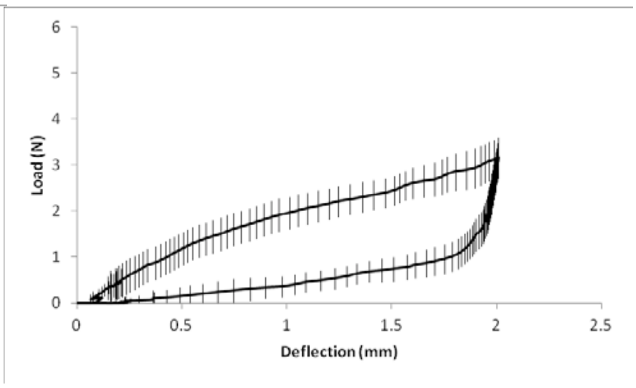


Figure 3.4b



The load-deflection plots of the Orthocare fully coated archwires are clearly different from the 3M uncoated control wires and from the Euroline labially coated archwires. A considerably reduced load was required to deflect the archwire to 1 or 2 mm (Figures 3.4a and 3.4b, respectively) demonstrated by a shallow loading gradient. Following unloading a relatively steep unloading gradient was observed and for wires deflected to 2 mm a plateau-like region was subsequently observed where the reduction in deflection was associated with a minimal reduction in force before returning to a zero deflection. In both Figures 3.4a and 3.4b a kink in the load-deflection data was observed at the onset of loading. Standard deviations were consistent throughout the loading and unloading profiles.

Figure 3.5 the mean load and the associated standard deviations (N) required to deflect Rhodium coated aesthetic archwire centrally to maximum deflection of 1 mm (Figure 3.5a) or 2 mm (Figure 3.5b) at a constant loading rate of 1mm/min. Once the peak deflection had been reached the unloading was undertaken at a constant rate of 1mm/min.

Figure 3.5a

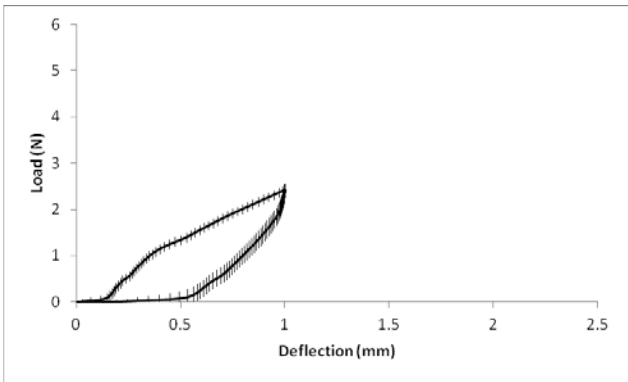
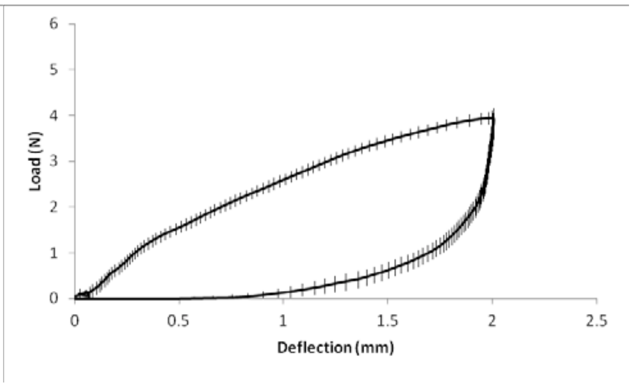


Figure 3.5b



The load-deflection plots of Rhodium coated aesthetic archwires were similar to the uncoated Sentalloy control (Figure 3.2) exhibiting strong consistency between samples illustrated by the narrow standard deviations in Figures 3.5a and b. In Figure 3.5a the loading curve does not gradually increase and instead there is an increased load between 0.25 mm and 0.5 mm and this could be attributed to possible operator induced effects introduced during ligation of the wire into the brackets. The subsequent behaviour showed a close similarity with the uncoated control. For both deflections at 1 and 2 mm a clear increasing reduction in load / deflection gradient is seen prior to the peak load being achieved and following unloading a sharp drop in load followed by a plateau region was obvious. Reduction to a negligible force application was observed when deflection was reduced to <50% of the peak deflection in Figures 3.5a and 3.5b.

3.2 IMPACT OF CLINICAL USE ON THE LOAD-DEFLECTION BEHAVIOUR OF COATED AND UNCOATED ARCHWIRES

The load-deflection behaviour of four types of 0.014” NiTi archwires retrieved after clinical use were characterised using 3 point wire bending tests (n=10 per group) and compared with unused equivalent wires.

3.2.1 3M uncoated archwires

Figure 3.6a (retrieved)

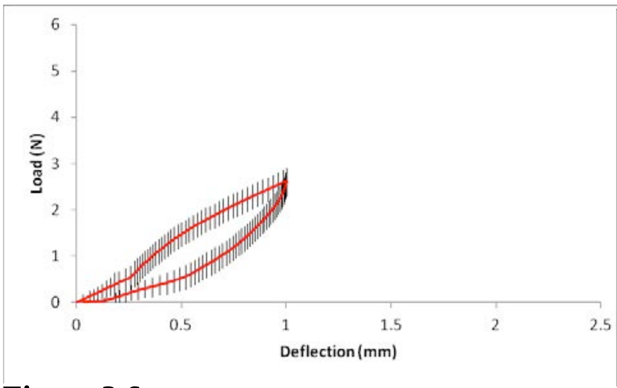


Figure 3.6b (retrieved)

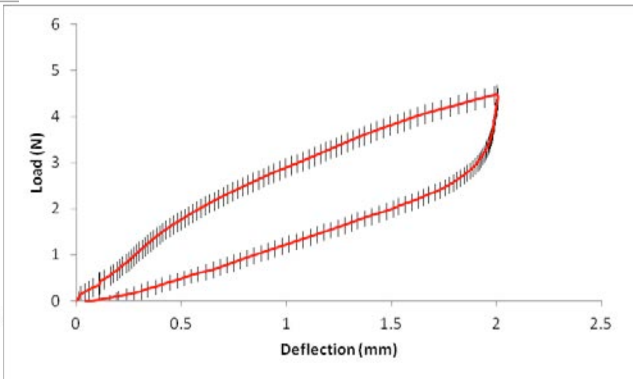


Figure 3.6c

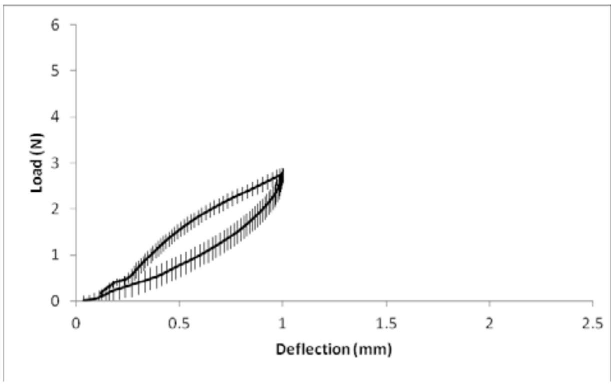


Figure 3.6d

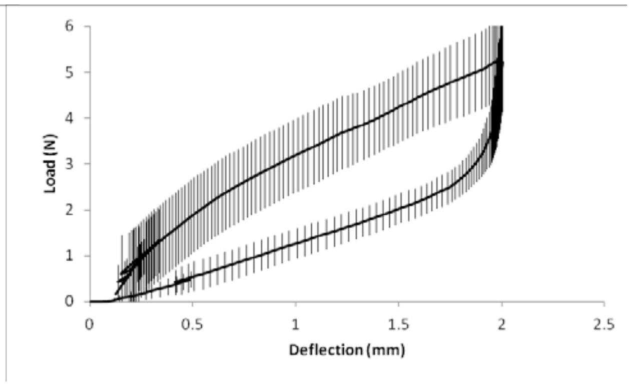


Figure 3.6a-d the load deflection behaviour of the 3M uncoated archwire after clinical usage (Figure 3.6a&b) compared with unused wires (Figures 3.6c&d) for loading and unloading measurements to a 1 mm (Figures 3.6a&c) and 2 mm (Figures 3.6&d) peak deflection. The plots highlight that the mean deflection profiles remain largely similar following clinical usuage. The key noticebale difference was a reduction in the standard deviations for the retrieved samples deflected to 2 mm when compared with the unused control.

3.2.2 Euroline labially coated aesthetic archwires

Figure 3.7a (retrieved)

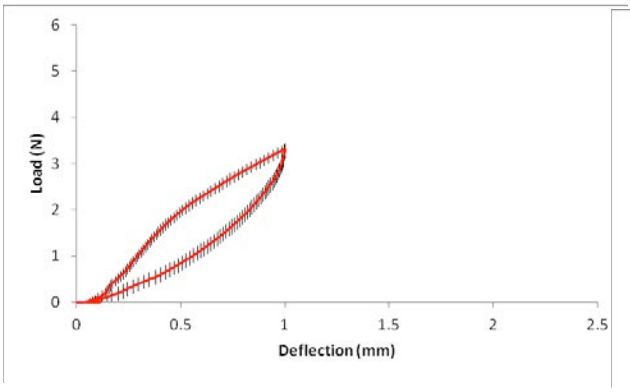


Figure 3.7b (retrieved)

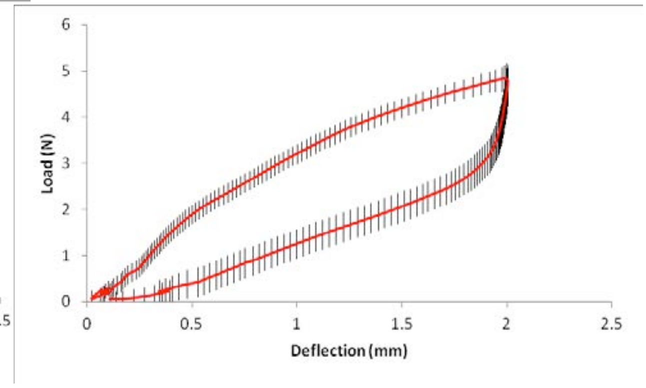


Figure 3.7c

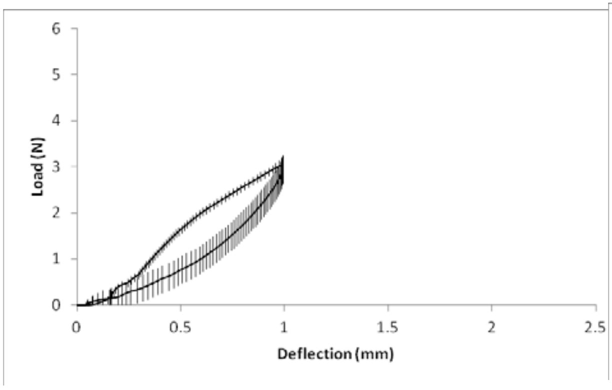


Figure 3.7d

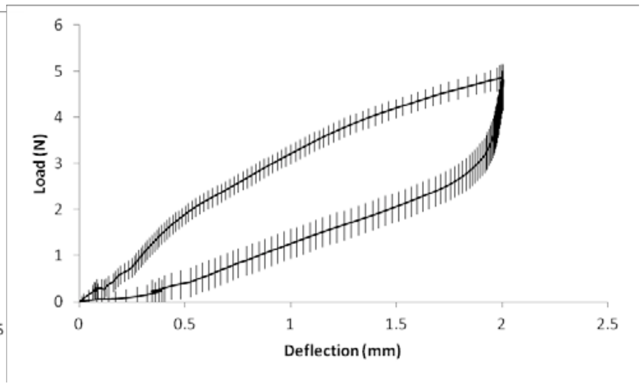


Figure 3.7a-d the load deflection behaviour of the Euroline partially coated archwire after clinical usage (Figure 3.7a&b) compared with unused wires (Figures 3.7c&d) for loading and unloading measurements to a 1 mm (Figures 3.7a&c) and 2 mm (Figures 3.7&d) peak deflection. The plots highlight that the mean deflection profiles remained largely similar following clinical usuage both in terms of the shape and for the 2 mm deflected speciemns in terms of the standard deviations of the measurements. For the 1 mm deflected specimens there was a reduced standard deviation in the unloading portion of the load-deflection curve for the retrievd archwires.

3.2.3 Orthocare fully coated aesthetic archwires

Figure 3.8a (retrieved)

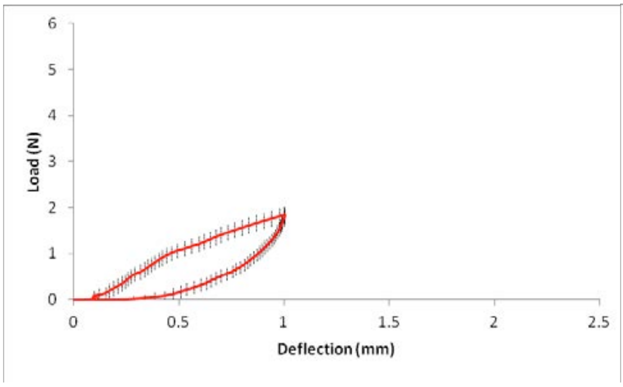


Figure 3.8b (retrieved)

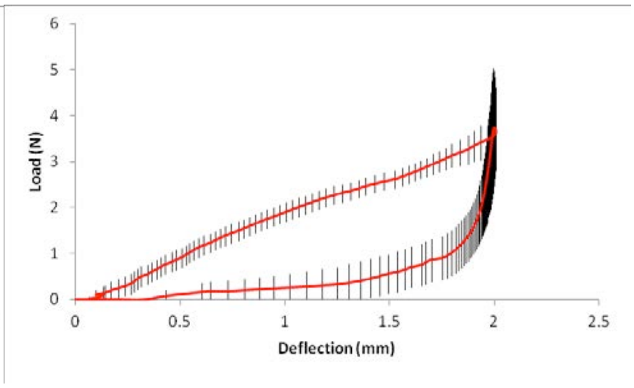


Figure 3.8c

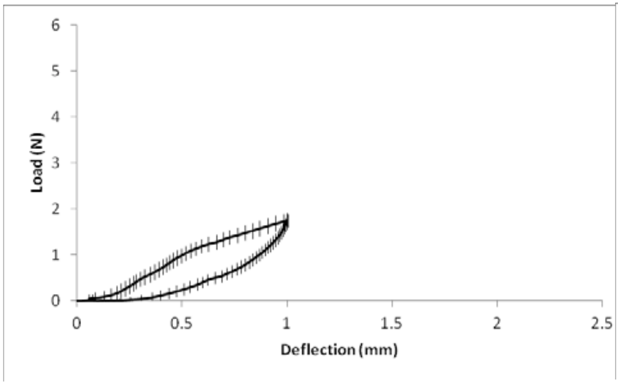


Figure 3.8d

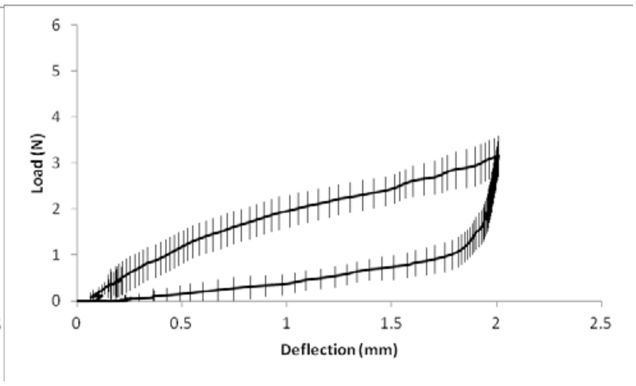


Figure 3.8a-d the load deflection behaviour of the Orthocare fully coated archwire after clinical usage (Figure 3.8a&b) compared with unused wires (Figures 3.8c&d) for loading and unloading measurements to a 1 mm (Figures 3.8a&c) and 2 mm (Figures 3.8&d) peak deflection. The plots highlight that the whilst the mean deflection profiles during loading remained largely similar, the peak force values were altered and the unloading profiles demonstrate modified gradients. For specimens deflected to 2 mm, increased variance in the measurements were obsrerved at values close to the peak deflection during loading and in the unloading profile.

3.2.4 Rhodium coated aesthetic archwires

For the Rhodium coated aesthetic archwires the unused control was a Sentalloy uncoated archwire which was considered to be the closest geometrical and compositional match to the experimental archwire.

Figure 3.9a (retrieved)

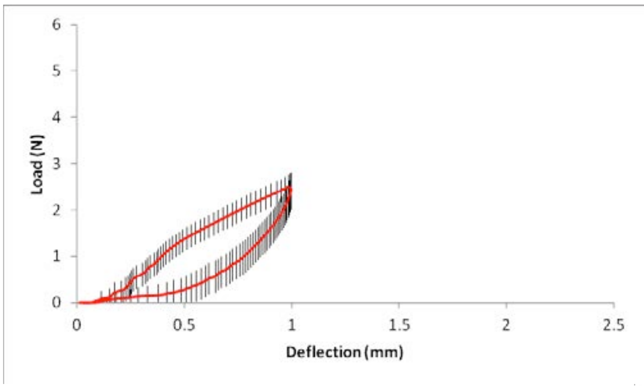


Figure 3.9b (retrieved)

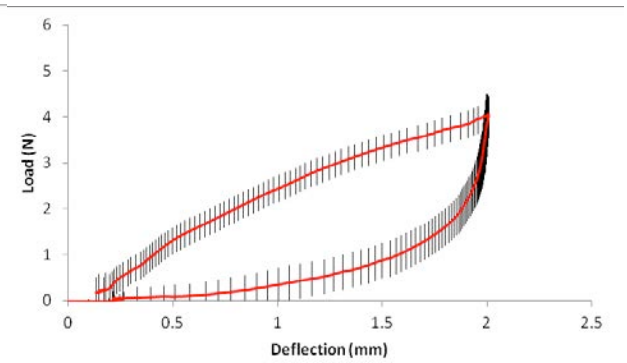


Figure 3.9c

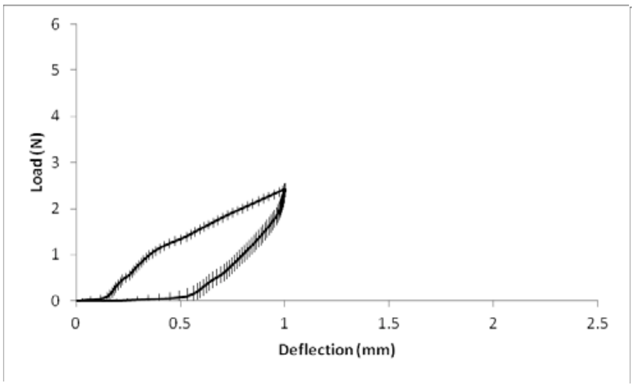


Figure 3.9d

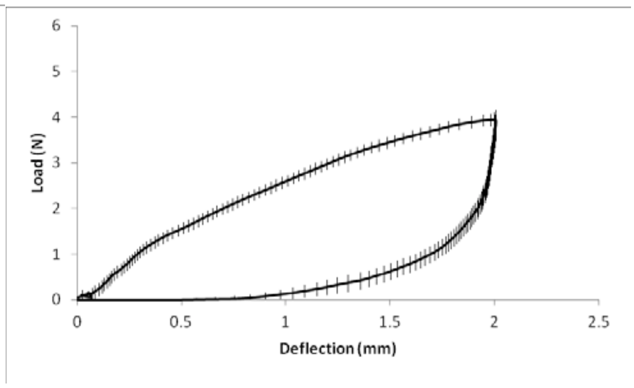


Figure 3.9a-d the load deflection behaviour of the Rhodium coated aestehtic archwire after clinical usage (Figure 3.9a&b) compared with unused Sentalloy wires (Figures 3.9c&d) for loading and unloading measuents to a 1 mm (Figures 3.9a&c) and 2 mm (Figures 3.9&d) peak deflection. The plots highlight that the whilst the mean deflection profiles during loading remained largely similar the variance in the measurements was increased following clinical

usage suggesting that there was a variable modification in the wire behaviour, presumably related to variability in the environmental exposure.

3.3 LOAD-DEFLECTION BEHAVIOUR OF DIFFERENT AESTHETIC ARCHWIRES

Figure 3.10 the mean load-deflection plots of the Rhodium coated, Euroline partially coated and Orthocare fully coated archwires measured using 3 point bending to a maximum deflection of 2 mm. Standard deviations are removed for clarity. The plots illustrate that there is significantly different behaviour observed between the three types of archwire. The Euroline partially coated wire demonstrates the steepest loading profile and peak load and least super-elastic behaviour during the unloading profile –evidenced by a clear lack of a plateau region.

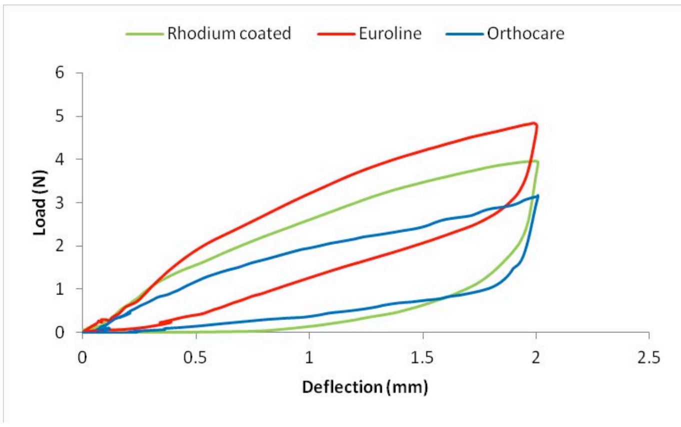
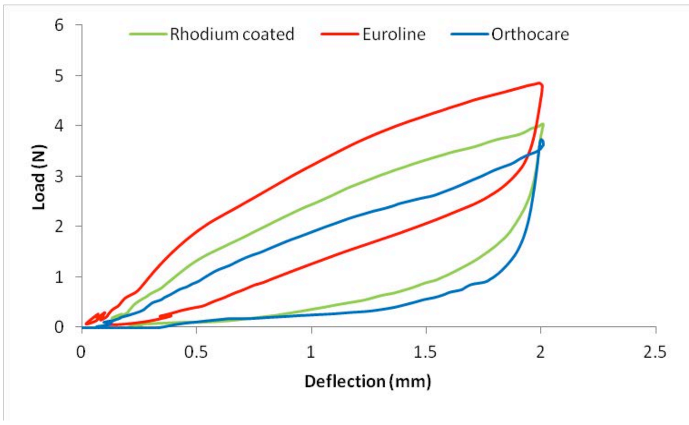


Figure 3.11 the comparative behaviour of the three aesethtic archwires following retrieval after clinical usage - standard deviations are removed for clarity.



3.4 IMPACT OF CLINCIAL USAGE ON MECHANICAL ARCHWIRE PARAMETERS

3.4.1 Unused archwires –deflected to 2mm

Table 3.1 summarises the results obtained for the 3 point wire bending tests carried out at a 2 mm deflection on the unused archwires. It includes the mean peak loads, the magnitude of the load at 1.5 mm deflection, 1 mm deflection during loading and the loading and unloading gradients related to the maximum deflection. Superscript annotations demonstrate statistical significance between dissimilar annotations (α =0.05)

Wire type	Mean Peak Load at 2 mm (N)	Mean Load at 1.5mm deflection (N)	Mean Load at 1mm deflection (N)	Mean Loading Gradient	R^2	Mean Unloading Gradient	R^2
3M Uncoated	5.24 (0.95)^{a}	2.04^{x}	1.27*	2.08	0.99	28.9	0.99
Sentalloy Uncoated	3.57 (0.34)^{b}	0.59^{y}	0.13***	1.6	0.99	38.2	0.99
Euroline Microcoated	4.84 (0.29)^{a}	2.08^{x}	1.25*	1.4	0.99	27.8	0.99
Orthocare Fully Coated	3.15 (0.04)^{c}	0.74^{y}	0.4**	1.34	0.99	23.2	0.99
Sentalloy Rhodium Coated	3.95 (0.16)^{b}	0.62^{y}	0.15***	1.13	0.99	33.8	0.99

3.4.2 Retrieved archwires –deflected to 2mm

Table 3.2 summarises the results obtained for the 3 point wire bending tests carried out at 2 mm deflection on the retrieved archwires. It includes the mean peak loads, the load at 1.5 mm deflection, 1 mm deflection and the loading and unloading gradients. Superscript annotations demonstrate statistical significance between dissimilar annotations (α =0.05)

Wire type	Mean Peak Load at 2 mm (N)	Mean Load at 1.5mm deflection (N)	Mean Load at 1mm deflection (N)	Loading Gradient	R^2	Unloading Gradient	R^2
3M Uncoated	4.47^{a} (0.21)	1.99^{x}	1.23^{*}	1.09	0.99	26.6	0.99
Euroline Microcoated	4.9^{a} (0.17)	1.81^{x}	1.21^{*}	2.14	0.99	35.1	0.99
Orthocare Fully Coated	3.37^{b} (1.18)	0.55^{y}	0.25^{**}	2.03	0.99	35.1	0.99
Sentalloy Rhodium Coated	3.87^{b} (0.61)	0.88^{y}	0.35^{**}	1.39	0.99	29.5	0.99

3.4.3 Unused archwires –deflected to 1mm

Table 3.3 summarises the results obtained for the 3 point wire bending tests carried out on unused archwires at 1 mm deflection. It includes the mean peak load at 1 mm, the load at 0.5 mm deflection and the loading and unloading gradients. Superscript annotations demonstrate statistical significance between dissimilar annotations (α =0.05).

Wire type	Peak Load at 1 mm (N)	Load at 0.5 mm deflection (N)	Loading Gradient	R^2	Unloading Gradient	R^2
3M Uncoated	3.73^{a} (0.2)	0.78^{x}	2.13	0.99	12.2	0.99
Sentalloy Uncoated	2.08^{b} (0.12)	0.22^{y}	1.8	0.99	6.9	0.99
Euroline Microcoated	3.3^{a} (0.12)	0.85^{x}	2.58	0.99	17.15	0.99
Orthocare Fully Coated	1.75^{b} (0.15)	0.25^{y}	1.36	0.99	5.22	0.99
Sentalloy Rhodium Coated	2.43^{b} (0.12)	0.08^{z}	2.01	0.99	14.9	0.99

3.4.4 Retrieved –deflected to 1mm

Table 3.4 summarises the results obtained for the 3 point wire bending tests carried out on retrieved archwires at 1 mm deflection. It includes the peak load at 1 mm, the load at 0.5 mm deflection and the loading and unloading gradients. Superscript annotations demonstrate statistical significance between dissimilar annotations (α =0.05).

Wire type	Peak Load at 1 mm (N)	Load at 0.5 mm deflection (N)	Loading Gradient	R^2	Unloading Gradient	R^2
3M Uncoated	2.63^{a} (0.26)	0.5^{a}	2.06	0.99	13.46	0.99
Euroline Microcoated	2.48^{b} (0.28)	0.28^{b}	2.1	0.99	13.3	0.99
Orthocare Fully Coated	3.03^{a} (0.19)	0.72^{a}	2.61	0.99	5.27	0.99
Sentalloy Rhodium Coated	1.85^{c} (0.16)	0.17^{c}	1.37	0.99	10.55	0.99

3.5 SURFACE ROUGHNESS RESULTS

Table 3.5 summarises the mean surface roughness (Ra-value [µm]) for wires as-used and in their unused condition. Sentalloy uncoated was not studied in this experiment for in-vivo use.

Wire type	Mean Ra -unused (µm)	Mean Ra –retrieved (µm)
3M Uncoated	0.06 (0.02)	0.07 (0.02)
Sentalloy Uncoated	0.10 (0.02)	not studied
Euroline Microcoated	0.4 (0.05)	0.71 (0.31)
Orthocare Fully Coated	0.34 (0.08)	1.37 (0.53)
Sentalloy Rhodium Coated	0.14 (0.03)	0.17 (0.05)

A two-way analysis of variance (ANOVA) demonstrated a significant impact of wire type ($p<0.01$) and of exposure (use or unused) ($p<0.01$) on the mean Ra-value. In addition there was a signifcinat interaction between wire type and exposure condition on the magnitude of the Ra-value (p=0.03). For the PTFE coated wires a signficinat increase in roughness was observed following oral exposure whilst changes in roughness were non-significnat for the uncoated alloys and the sentalloy coated alloys.

3.6 SEM RESULTS

SEM was undertaken to qualitatively study modifications to the surface topology of the archwires after clinical usage.

Figure 3.12a SEM images of a 3M uncoated, unused archwire.

The scanning electron micrographs demonstrate that the archwire has a smooth surface with areas of pitting visible. At increased magnification lines and grooves consistent with the manufacturing process are apparent parallel to the long axis of the archwire.

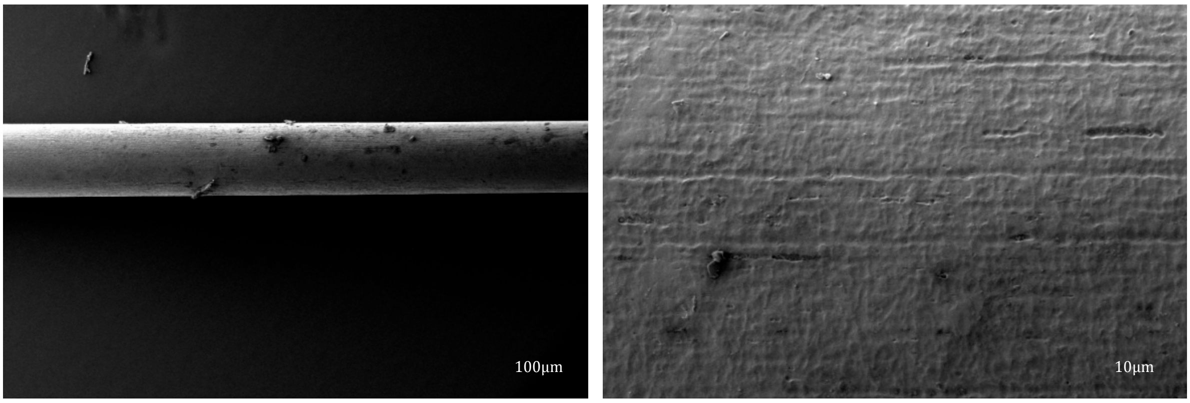


Figure 3.12b SEM images of a 3M uncoated retrieved archwire.

The retrieved uncoated archwire has a similar appearance to the unused archwire, with little evidence of deterioration. The lines and grooves on the surface of the archwire are visible.

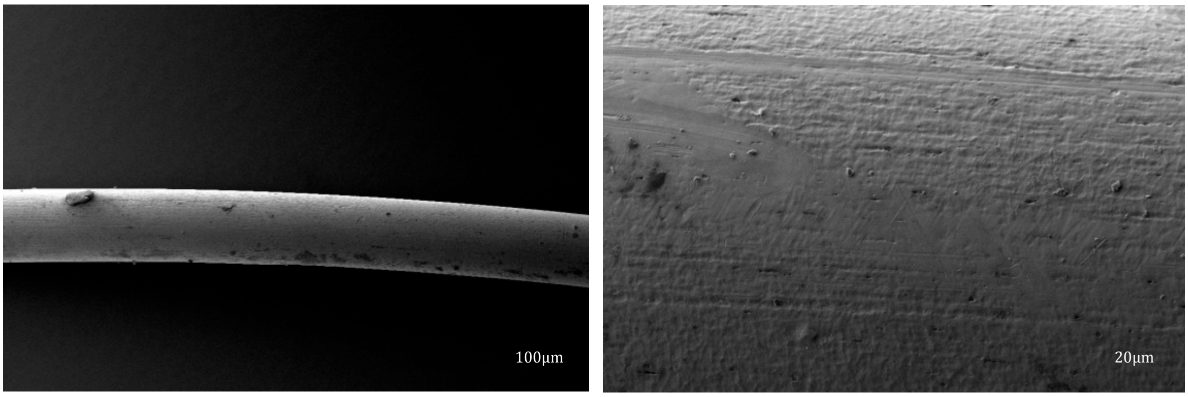


Figure 3.13a SEM images of a Sentalloy uncoated, unused archwire

The sentalloy uncoated archwire has a smooth profile with minor irregularity visible at x1000 magnificaintion. There are however no obvious defects evident on the archwire surface.

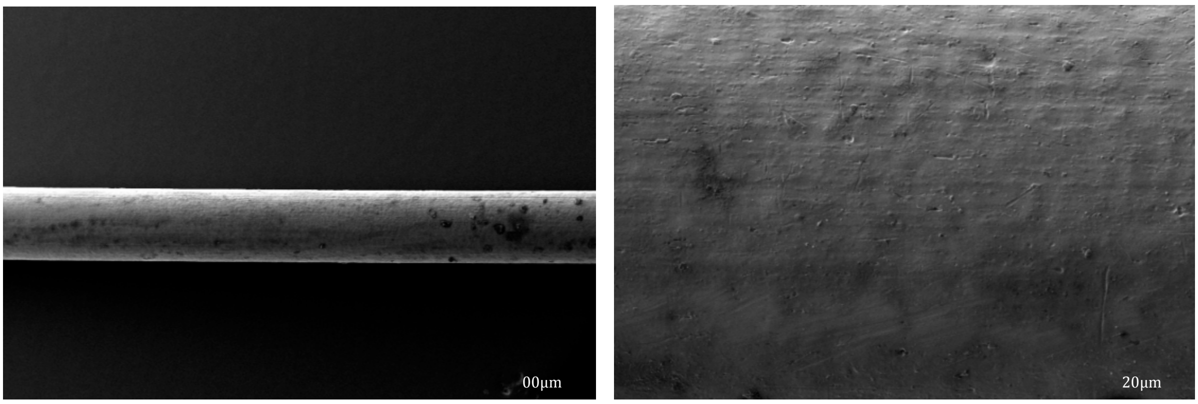


Figure 3.14a SEM images of a Euroline labially coated, unused archwire

A clearly visible coating layer is seen on the archwire, with areas of roughness adjacent to this layer. Grooves are visible on the surface of the coating at increased magnification.

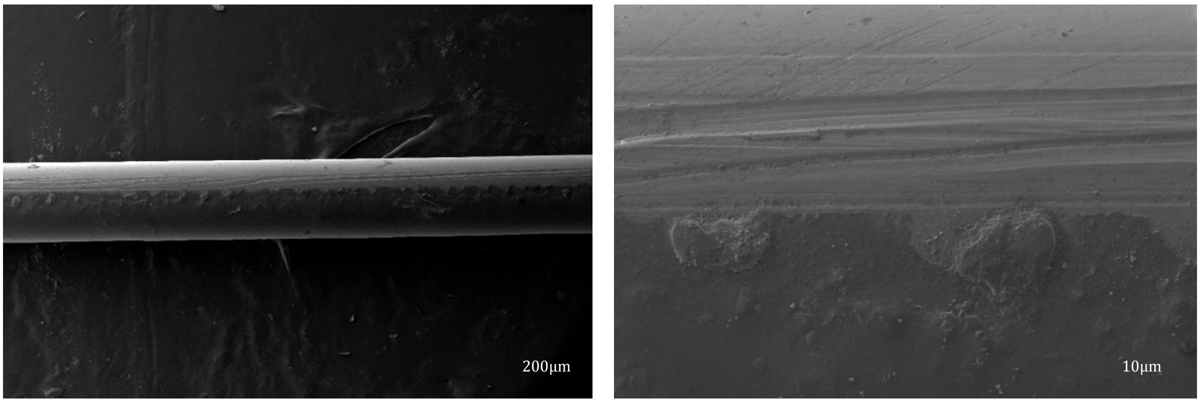


Figure 3.14b SEM images of a Euroline labially coated, retrieved archwire

Important morphological variations are apparent when the retrieved archwire is compared with the unused archwire. Large areas of delamination are clearly seen on the surface of the archwire and exposed areas of archwire are visible as a result of intraoral aging. The remaining coating material looks rough in appearance.

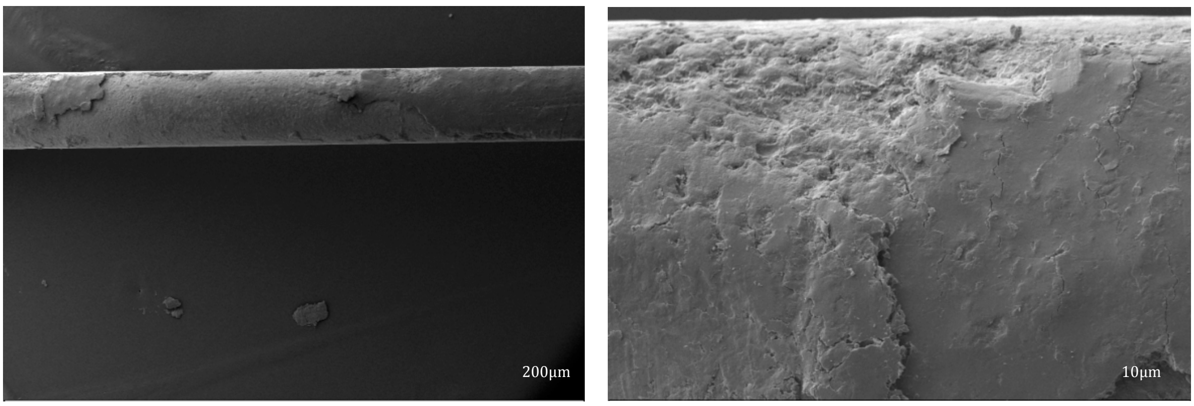


Figure 3.15a SEM images of a Orthocare fully coated, retrieved archwire

The unused archwire has a smooth profile but there is an area of delamination and evidence of the coating peeling off. This may be attributed to handling of the archwire while preparing it for viewing under the scanning electron microscope. There are also numerous pits on the surface of the archwire, giving it an uneven profile.

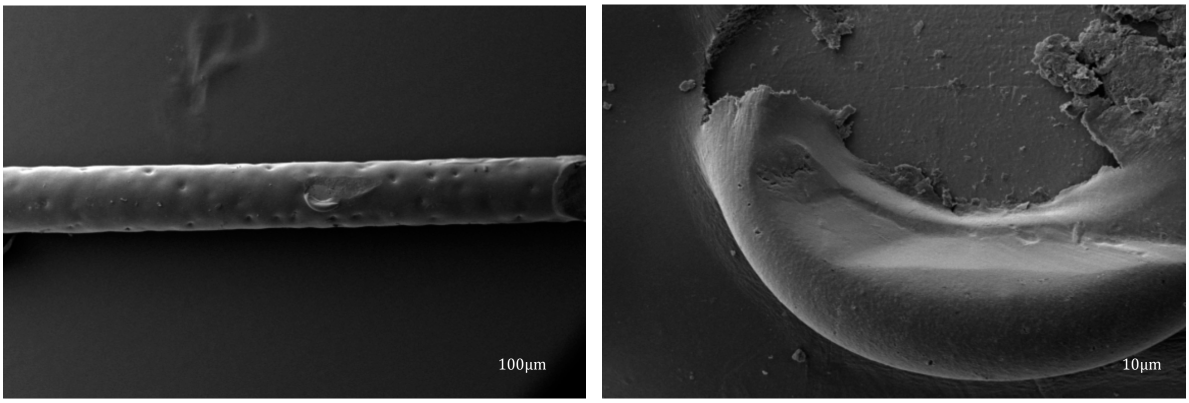


Figure 3.15b SEM images of a Orthocare fully coated, retrieved archwire

Clearly visible macroscopic changes on the surface structure of the retrieved archwire are demonstrated. There are large areas of missing coating and roughness on the surface of the wire. This would clearly affect the aesthetic value of the archwire.

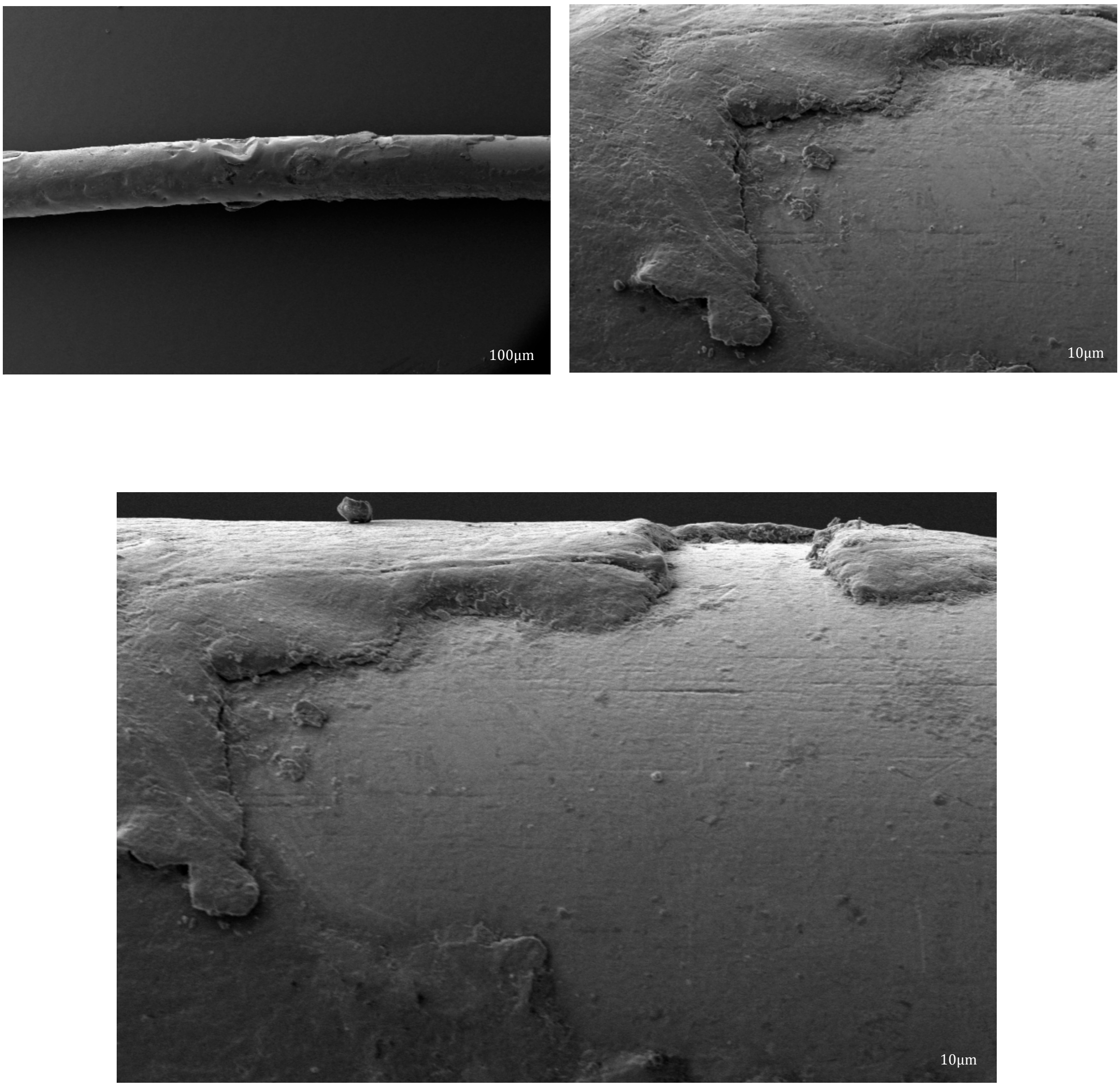


Figure 3.16a SEM images of a Sentalloy rhodium coated, unused archwire

Although there is no obvious coating seen on the surface of the archwire at the magnifications employed, the surface does appear coarse.

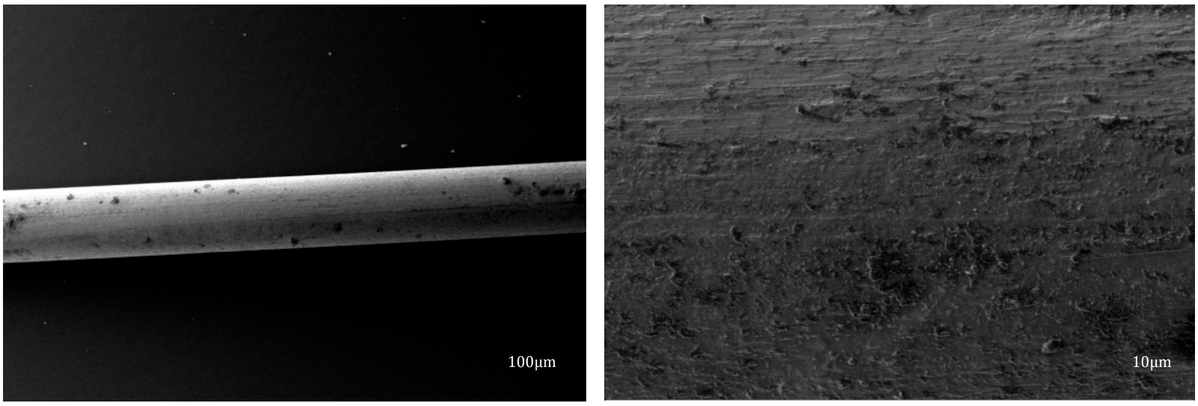
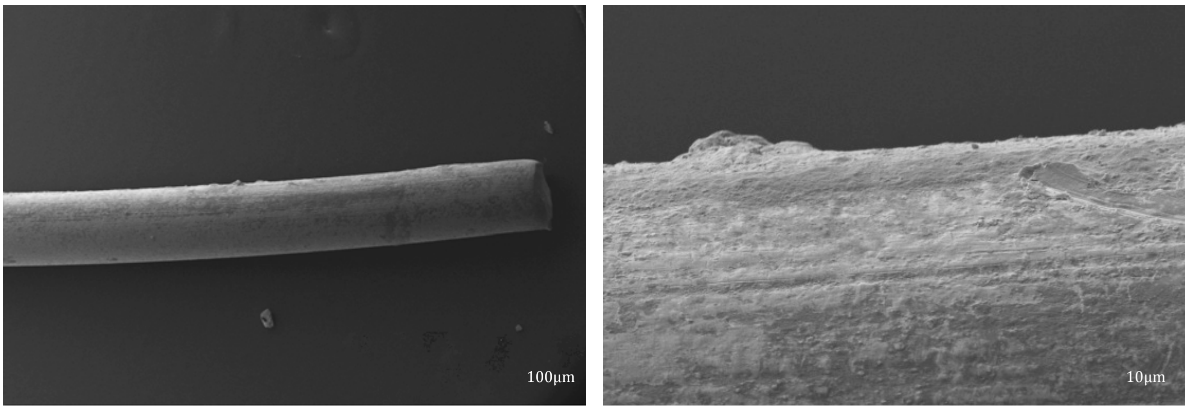


Figure 3.16b SEM images of a Sentalloy rhodium coated, retrieved archwire

Although the surface of the retrieved archwires appears rougher then the unused archwire there is not a great difference between the unused and retrieved archwires.



Chapter 4 – Discussion

The purpose of this study was to compare the mechanical and surface properties of uncoated and coated ‘aesthetic’ archwires before and after clinical usage. 3 point wire bending tests were conducted to characterise the load-deflection relationships generated by loading and unloading a ligated archwire. The upper loading curve represents the force needed to engage the archwire in the orthodontic bracket and the lower, unloading curve represents the force delivered during the initial aligning and levelling stage of treatment (Segner *et al.,* 1995). The experimental set-up is only an approximation of the complex situation intra-orally but allows the wires to be studied systematically under similar conditions and allows the effects of friction and binding at the wire-bracket interface to be considered alongside the stress-strain behaviour of the wire itself. Wire behaviour at two deflections was considered where a maximum 1 mm deflection was thought to represent a minimally activated wire and 2 mm deflection a moderately deflected wire.

4.1 IMPACT OF CLINICAL USE ON THE BEHAVIOUR OF COATED AND UNCOATED ARCHWIRES

4.1.1 Load-deflection behaviour

For many of the experimental measurements obtained in the current study there was considerable variability between repeat samples of the same wire type. The findings demonstrate that the impact of clinical usage on the behaviour of coated and uncoated archwires is variable. When comparing the load-deflection of the unused and retrieved archwires it was evident that the 3M uncoated archwires behaved similarly at a 1 mm maximum deflection but increased standard deviations were evident when the wires were deflected to 2mm. In addition following unloading the unused 3M archwire did not deflect back to a zero reading. These observations can be explained by either permanent distortion of the archwire (deformation beyond the yield point) occurring or that there has been some sliding of the archwire at the ligation sites. Given a maximum deflection of 2 mm was introduced the latter mechanism was considered more likely and therefore is considered in the context of the behaviour of all of the experimental measurements. The behaviour of Orthocare fully coated archwires were largely similar and interestingly the loading curves exhibited small fluctuations in the measured load as the wire was been deflected. This observation was attributed to inconsistencies in the friction between the wire and the ligation/bracket site and was only observed in the coated archwire samples. The only notable difference between unused and retrieved Orthocare fully coated archwires when deflected to 2 mm were the steeper unloading curves with the retrieved archwires and greater standard deviations suggesting the retrieved fully coated wires lose force rapidly once the tooth starts to move and that the force delivered by the wires is more variable.

Load deflection curves for the rhodium coated archwires demonstrated that when the wires were deflected to 1mm both the unused and retrieved archwires exhibited steep loading and unloading profiles which is indicative of little sensitivity in force application and rapid loss of useable force following small amounts of tooth movement. Similarly to the Orthocare fully coated archwires there was some unevenness in the initial loading curve which was attributed to wire movement at the ligation site. When deflected to both 1 mm and 2 mm the retrieved rhodium coated archwires have greater standard deviations than the unused wires, again suggesting that the behaviour of retrieved archwires is less predictable after 6 weeks of clinical use. This in turn can be related to either modification of the surface (surface deterioration or development of surface biofilm) or to alloy modification itself. The latter although less likely cannot be discarded as the control comparisons for the Rhodium coated archwires were an unused Sentalloy equivalent.

The load-deflection behaviour of the Euroline labially coated archwires was similar for unused and retrieved samples. Interestingly when the wires were deflected to 1 mm steep loading and unloading curves were evident with no obvious plateaus and signs of stress induced phase transformations. The behaviour could be typical of the wire itself or could represent a modification of the stress-strain behaviour of the wire by the coating itself. The Euroline archwires were labially coated and the force applied during 3 point bending was on the contralateral surface putting the coating into tension during the 3 point bending test. When a greater deflection of 2mm was induced the wire behaved in a more typical manner of NiTi archwires. As seen with previous archwires there are greater standard deviations in the overall behaviour of the retrieved archwires which is again indicative of some modification to the coating and/or wire which is sensitive to the specific environmental exposure.

4.1.2 Engagement and active forces

No differences in the force needed to engage the Orthocare fully coated NiTi archwires before and after 6 weeks of clinical use were observed despite obvious degradation and delamination of the coating. The Euroline labially coated archwires did however have a reduced peak load at 1mm deflection after 6 weeks of usage, suggesting that reduced force is needed to engage the archwire after clinical use. This suggests that an intact labial coating may contribute some stiffness to the wire which is subsequently lost after a period of clinical usage. From a clinical perspective the predictability of this behaviour change in the context of when it happens is uncertain. Alternatively the reduced load to achieve a 1mm deflection may also be related to sliding at the ligated brackets but as the load deflection plots tended to return to the unloaded origin following the loading cycle this was considered to be less likely. The 3M uncoated archwires also produced statistically significant reduced loading forces at 2 mm deflection following clinical usage.

Retrieved Orthocare fully coated archwires produced lower unloading forces at 1.5 mm, 1 mm and 0.5 mm deflection compared with as-received fully coated archwires. Similarly, retrieved Euroline labially coated archwires generated lower unloading forces at 1.5 mm and 0.5 mm deflection. The most likely reason for this is that increased friction is created between the irregular surface of the coated archwire and the bracket. Defects on the surface of the archwires where the edges of the brackets and the archwires contact may impede the wire sliding through the brackets during the 3 point wire bending tests. A previous study by Elayyan *et al* (2008) deflected unused and retrieved coated archwires by 2 mm and 4 mm and they also found that retrieved coated archwires deflected to 2 mm produced lower unloading forces whereas when the wires were deflected by 4 mm they generated unloading force values of zero. After examining the archwires they concluded that they were 'jammed' in place and prevented from sliding through the brackets due to damage of the coating and increased friction. This 'jamming' effect resulted in the wires remaining static as the load was released. Bradley *et al* (2013) in contrast found that retrieved aesthetic archwires had significantly higher stiffness and force values than the same wires before use and concluded that as the coating was lost during clinical use the archwires behaved more like their uncoated counterparts.

The unloading gradients of the retrieved Orthocare fully coated archwires deflected to 1mm was significantly increased suggesting that the force delivered to the teeth reduces rapidly following initial movement and alignment of the teeth. When comparing the peak load of the archwires deflected to 1 mm and 2 mm before and after use differences were observed. The difference in peak load of the Euroline archwires deflected to 2 mm before and after use was not statistically significant whereas the Euroline archwires produced reduced forces after 6 weeks of clinical use when deflected to just 1 mm. When the archwires are deflected to 1 mm some of the movement of the wire will be the stretching of the elastomeric modules and static friction may not be overcome. When the archwires are deflected to 2 mm however, they are more likely to displace and overcome static friction irrespective of coating state and hence you have less of a difference in the peak loads before and after use.

4.2 IMPACT OF COATING ON THE BEHAVIOUR OF NICKEL-TITANIUM ARCHWIRES

3 point wire bending tests were also used to determine the tooth-moving properties of the coated and uncoated archwires. The loading and unloading curves of the Sentalloy rhodium coated archwires are steeper than the uncoated archwires. Both wires have steep unloading curves which suggest that once the tooth starts moving the force delivered is rapidly reduced. The rhodium coated archwire deflected to 2 mm performed better than the uncoated Sentalloy archwire, producing a higher unloading response. The wires also produced higher loading forces when deflected to 1 mm. This observation may be attributed to minor deviations in the individual dimensions of the archwires and consequential variations in the load deflection properties of the uncoated and rhodium coated archwires. It may also be that the manufacturers use different stock archwires for rhodium coated wires as opposed to simply coating the original archwires. There were greater standard deviations observed for the uncoated Sentalloy archwires suggesting these wires behaviour are more unpredictable.

The load deflection curves of the 3M uncoated, Euroline labially coated and Orthocare fully coated archwires when deflected to 1 mm are similar suggesting that at small deflections the coated archwires behaviour initially in a similar way to the uncoated counterparts. However when deflected to 2 mm differences in the wire behaviour became apparent. It is likely at 1mm deflection that magnitude of wire movement at the ligated bracket site will be small however at 2mm deflections there is evidence from inconsistencies in the loading profiles that some translational movement of the wire through the bracket must occur. Associated with this was considerable variability between repeat samples leading to large measurement standard deviations. The 3M uncoated archwire deflected to 2 mm demonstrated particular variability which may be explained by operator variability in securing the wire to the experimental jig or by intrinsic variability in the surface finish of the wire provided ‘as manufactured’.

There was no statistically significant difference noted between the loading and unloading forces of the 3M uncoated, Orthocare fully coated and Euroline labially coated archwires when deflected to 2 mm and 1 mm. Previous studies that have found lower unloading forces for the coated archwires suggested this was most likely due to the coated wires having smaller cross-sectional dimensions. A recent study by Da Silva *et al* (2013) examined the inner alloy core dimensions of a variety of coated archwires and found no significant difference in the cross-sectional diameter of the coated and uncoated archwires for one manufacturer (TP Orthodontics: Aesthetic Shiny Bright and non-coated Shiny Bright archwires). Subsequently, Washington *et al* (2014) used a Nikon Profile Projector to examine the dimensions of archwires at 100x magnification and found similar results for TP and GH wires. Their study also found the archwires produced by manufacturers with approximately the same dimensions as their uncoated counterparts exhibited similar load responses as the uncoated archwires. As the thickness of the coated wires used in this study was only between 0.002” - 0.005” the archwire dimensions of the coated and uncoated wires may be similar, resulting in coated wires that perform as well as their uncoated counterparts. The deflection of the archwires by 1 mm or 2 mm did not impact on the behaviour of the wires when subjected to 3 point wire bending tests.

Ellayan *et al* (2008) used self-ligating brackets when testing the mechanical properties of the coated and uncoated archwires before and after clinical use and found similar loading and unloading forces. This may be due to the reduced friction associated with self-ligating brackets. They may not be as affected by the delamination and increased surface roughness as conventional brackets. In addition, Eliades *et al* (2000) examined the surface characteristics of coated archwires after 1-6 months of clinical use and found that the method of ligation i.e. stainless steel ligatures versus elastomeric modules, did not influence the characteristics of biofilm maturation or the aging pattern.

Comparisons of the force levels generated by coated archwires in previous studies cannot be made as there are no other studies that have used 0.014” coated archwires in 3 point wire bending tests, with round 0.016” and 0.018”x 0.025” rectangular archwires being more commonly used for testing. In addition there are other variables such as method of ligation, length of archwire span, the size of the loading instrument, loading speed and the direction of testing that can alter the results between studies.

4.3 SURFACE CHARACTERISTICS OF UNUSED AND RETRIEVED COATED AND COATED ARCHWIRES

SEM was carried out on unused and retrieved archwires under 100, 1000 and 1,500 magnification to analyse the surfaces of the wires. The uncoated 3M and Sentalloy archwires showed the typical surface characteristics of superelastic archwires and had smooth surfaces with some evidence of the manufacturing process visible and no obvious defects. The PTFE coated Orthocare and Euroline unused archwires showed grooves and pitting and the rhodium coated archwires had a smooth profile with minor irregularities visible at x1000 magnification. After 6 weeks of clinical use the PTFE coated archwires demonstrated a considerable amount of deterioration, this is consistent with the results of previous studies (Neumann *et al,* 2002; Elayyan *et al* 2008; Bradley *et al,* 2013). The loss of coating after a mean period of 21-55 days has been reported to be between 25% to 72.9% (Elayyan *et al* 2008; Bradley *et al,* 2013; Da Silva *et al*, 2013). Although there was a loss of coating the NiTi wires beneath the coating appeared smooth and defect-free. The deterioration in the surface PTFE coating was associated with a concomitant significant increase in the measured surface rough (mean Ra values) which would clinically impact on friction within the bracket slots.

The delamination and deterioration of the coating of the archwires may affect the aesthetic properties of the wires and consequently patient satisfaction. There are also a number of clinical consequences that may arise as a result of the loss of the coating. Plaque may accumulate more readily around areas of delamination, making oral hygiene more challenging in these areas. Although the wire beneath where the coating appeared without defects, it may in time become more susceptible to corrosion. The sliding properties of the archwires are also likely to reduce significantly as the bracket edges become entrapped inside the defects although further clarification of the implication of surface roughness on friction is needed. Excessive wear was noted where the wires were engaged with the brackets. This may be as a result of the force used to ligate the wires with elastomeric modules or it may be attributed to the ploughing effect of the coating during sliding of the wires on the brackets.

Given the delamination and deterioration of the coating following clinical use it is probable that these wires may not satisfy the aesthetic demands of the patients that request aesthetic appliances. Bradley *et al* (2013) found that at least half of patients were aware of texture and colour changes to PTFE coated archwires over time. There have been no other studies that have looked at patient satisfaction with aesthetic archwires. Improvements to the coating technique or the development of clinically acceptable composite archwires in the future may overcome the limitations of aesthetic archwires.

4.4 LIMITATIONS OF THE STUDY

The results of this study are limited to the materials tested. Therefore they can only be applied to the specific 0.014" round NiTi archwires used with pre-adjusted edgewise brackets and conventional elastomeric ligation. Rectangular and stainless steel archwires may behave differently. In addition the archwires were deflected to 1 mm and 2 mm and the archwires may behave differently if they are deflected to greater than 2 mm or indeed less than 1 mm. As previously discussed the method of ligation may also influence the behaviour of the superelastic coated and uncoated archwires.

Retrieval analysis has gained interest in recent years as it provides information on the performance of dental materials in the environment in which they are intended to operate and it also aids in assessing any alterations of the properties of the materials that occur during their clinical use. The sequence in the changes to the properties of the archwires cannot be provided by retrieval analysis and this is a drawback of this form of mechanical testing. The limitations of using raw data from the mechanical testing of archwires should also be acknowledged. Although 3 point wire bending tests are the most commonly used tests for assessing the performance of archwires the situation in the mouth may be different and statistically significant values may not translate into altered clinical performance.

There are many factors which could have influenced the integrity of the coated archwires and thus the results, such as differences in oral hygiene habits, the use of fluoride mouthwash and toothpaste, oral pH, biological factors and the intraoral exposure period. Although the SEM revealed significant areas of delamination of the coating on the archwires it may have been useful to quantify the amount of remaining coating. This has been carried out in three previous studies using different methods. Elayyan *et al* (2008) transferred digital images of the archwires into a computer and used an Image®J program and excel to calculate the remaining coating. Bradley *et al* (2013) scanned images of the wires, imported them though Matlab and used *csvwrite* to apply three numerical values for each pixel. The percentage of the wire was divided by the sum of the percentage of the coating and the wire to calculate an evaluation percentage. Da Silva *et al* (2013) used Image ProPlus 4.5 software to calculate the amount of coating remaining on the retrieved archwires. However, when relating coating loss to load deflection data it is essential to recognise that quantification of coating deterioration may not always be relevant as it is likely to be localised patterns of deterioration at zones of direct interaction with the bracket or ligature which are responsible for the variability in behaviour observed. A further weakness of this study is that the 3 point wire bending tests were carried out at 24 °C instead of 36 °C. This reduced temperature may lead to reduced load response values as the superelastic deformation behaviour of nickel titanium archwires is highly dependent on deformation temperature. The stresses for superelastic loading and unloading increase with increasing temperature.

Chapter 5 – Conclusion

The purpose of this study was to examine the mechanical and surface properties of coated ‘aesthetic’ archwires before and after clinical use.

1. The findings demonstrate that the impact of clinical usage on the behaviour of coated archwires is variable. The archwires behaviour after 6 weeks of clinical use is more unpredictable than that of unused archwires with greater variability between the repeat samples. This can be related to either modification of the surface (surface deterioration or development of surface biofilm) or to alloy modification itself. In addition the retrieved PTFE coated archwires produced lower unloading forces than unused archwires which may be due to increased friction between the irregular surface of the coated archwire and the bracket.

2. The force generated by the uncoated and PTFE coated archwires is comparable. There was no statistically significant difference noted between the loading and unloading forces of the 3M uncoated, Orthocare fully coated and Euroline labially coated archwires when deflected to 2mm and 1mm. Therefore when selecting the PTFE coated archwires used in this study operators can expect the teeth to move at comparable rate to their uncoated counterparts.

3. The coated archwires undergo delamination and degradation of the coating after 6 weeks of clinical use which may affect the aesthetic properties of the wires and consequently patient satisfaction. It may also make oral hygiene more challenging in these areas and patients should be warned about this.

5.1 FUTURE WORK

Future studies could include coated NiTi and stainless steel archwires of varying cross sectional diameters, including both round and rectangular archwires. It would also be useful to test the wires at greater deflections such as 3 mm and 4 mm using the same 3 point wire bending tests. This would give a greater picture of how the wires perform when there is greater irregularity of the teeth.

Given the delamination and deterioration of the coating of the archwires used in this study it might be helpful for future studies to measure the amount of coating remaining on the archwires after clinical use. In addition obtaining feedback from patients on their satisfaction with the coated archwires before and after clinical use would also be beneficial.

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Appendix 1A - Email to NRES to clarify the need for ethical approval

[redacted]

Appendix 1B - Email to clarify the need for ethical approval

[redacted]

Appendix 1C - Email from NRES confirming that ethical approval is not sought for this research.