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Mechanical Properties of Orthodontic Coated Stainless Steel Arch Wires
(An in vitro study)

A Thesis Submitted to the Council of the College of Dentistry, University of Baghdad in Partial Fulfillment of Requirements for the Degree of Master of Science in Orthodontics

By
Ahmed Abdulhussain Abbas
B.D.S.

Supervised by
Prof. Dr. Akram Faisal Alhuwaizi
B.D.S., M.Sc., Ph.D.

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Dedication

I would to dedicate this thesis

- *To the first teachers who taught me the letters of life with spiritual love and care "My lovely mother and great father"*
- *To my dear brothers and lovely sister who greatly support and encourage me to finish this study successfully*

Ahmed

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Abstract

The increased demands of aesthetic orthodontics have led the manufacturers to introduce orthodontic aesthetic wires, balancing between aesthetic requirements and improved mechanical properties.

This study was established to evaluate the exact dimensions, thickness of coating layer and mechanical properties of coated orthodontic arch wires from seven companies with two wire dimensions 0.016''×0.022'' and 0.019''×0.025'' making a total number of 238 coated wire segments and 34 uncoated wire segments as control samples.

Total wire dimension, inner core dimension and thickness of coating layer was measured using a metallurgical light microscope.

A universal testing machine was used with two types of tests: a three-point bending test to determine different mechanical properties including (maximum deflection force, yield strength, modulus of elasticity and spring-back) and frictional resistance test to evaluate the maximum static frictional forces produced by each type of arch wires. ANOVA and HSD tests were used to statistically analyze the results.

Results showed that there are relatively high differences between measured wire dimensions and the stated dimension among coated arch wires. Labially coated arch wires from DB and fully coated wired from DANY demonstrate higher maximum deflection force, higher yield strength and higher modulus of elasticity, labially coated wires from RMO and TP demonstrate mechanical properties nearly similar to the control wires, while fully coated wires from GH, Highland and Hubit showed the least values of mechanical properties. Spring-back property was different among wire types.

High frictional resistance force has been confirmed with the labially coated wires from DB, RMO and TP while the least values were with the fully coated wires from Hubit owing to differences in the total wire dimensions, thickness of coating layer, modulus of elasticity and surface roughness.

It can be concluded that inner alloy dimension, thickness of coating layer, alloy composition and physical properties of coating layer play a major role in the mechanical properties of coated stainless steel wires.

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List of Abbreviations

Abbreviation	Full text
%	Percentage
®	Registered Trademark
°C	Degree Centigrade
''	Inch
ANOVA	Analysis of Variance
β-Ti	Beta-Titanium
Co-Cr	Cobalt-Chromium
d.f	Degree of freedom
etc.	et cetera
Fig	Figure
FRPC	Fiber-Reinforced Polymer Composite
GPa	Giga Pascal
Gm	Gram
ISO	International Organization for Standardization
LSD	Least significant difference test
MPa	Mega Pascal
Mil	One thousandth of an inch
mm	Millimeter
min	Minute
Min	Minimum
Max	Maximum
N	Newton
NiTi	Nickel titanium
p-value	Probability value
PTFE	Polytetraphloroethylene
RMO	Rocky mountain orthodontics ®

SD	Standard Deviation
SRPC	Self-Reinforced Polymer Composite
TMA	Titanium-molybdenum alloy
USA	United States of America
UK	United Kingdom

Introduction

Introduction

Since orthodontic treatments extend over a number of months, the aesthetic appearance of the appliance is valued by patients as an important factor worthy of consideration. In order to improve the aesthetics, orthodontic wires as well as brackets, tooth-colored coatings have been developed **(Kim *et al*. 2014)**.

Although, plastic or ceramic esthetic brackets are generally used in aesthetic orthodontics, most orthodontic arch wires are made from metal with a coating material (**Lopes *et al*., 2012)**.

Four different types of materials have been used for coating of metal arch wires such as Teflon, Epoxy, polymer and rhodium coated material **(Gopal, 2010; Ryu, 2015).**

The coating of orthodontic arch wires is likely to influence their characteristics including thickness, surface roughness and mechanical and frictional properties **(Totino *et al*., 2014; Kim, 2014 Rudge *et al*., 2015; Washington; 2015; Choi, 2015).**

Manufacturers of orthodontic materials are currently investing in the search for the ideal wire coating that combine both aesthetics and mechanical efficiency. The different types of coatings can change some wire properties, so addressing to what extent these changes have occurred is not yet well know **(Lim *et al*.,1994; Kusy, 1997; Elayyan *et al*., 2008 and Elayyan *et al*., 2010).**

Proper selection and understanding of the biomechanical requirement of each case requires proper characterization studies on arch wire alloys **(Goldberg *et al*., 2010).**

Several new aesthetic wires have been introduced continuously into clinical practice, and there has been few published literature concerning the detailed mechanical properties of coated stainless steel wires **(Tang, 2017).**

The present study has been established to determine and compare the mechanical properties for different aesthetic arch wires coated with different materials from different manufacturer and depending on standardized specifications.

Aims of the study

This in vitro study was designed and accomplished to measure, evaluate and compare different characteristics of coated stainless steel arch wires among different companies including the exact total dimensions, inner core dimensions and thickness of coating layer and their effects on the values of different mechanical properties:

- Maximum deflection force
- yield strength
- modulus of elasticity
- spring-back
- frictional resistance force

Chapter One

Review of Literature

Chapter One

Review of Literature

1.1 Orthodontic arch wires

1.1.1 Introduction

An appropriate orthodontic tooth movement can best be attained by application of optimal force system which has moderate to low force magnitude and result in rapid and relatively painless tooth movement with minimal damage to surrounding tissues **(Agwarwal *et al.*, 2011)**.

The biomechanical force system formed during orthodontic treatment depends on the appliance design and the mechanical and physical properties of the arch wires being used during treatment **(Sander *et al.*, 2009)**.

A several important factors should be considered during selection of arch wires for a certain stage of orthodontic treatment including: amount of force desired, yield stress, modulus of elasticity, and simplicity of soldering and welding **(Daems *et al*, 2009).** Understanding the basic material characteristics became essential for selecting wires for use in the treatment **(Gatto *et al.*, 2011).**

A light continuous force is always desired during orthodontic tooth movement, such forces may reduce patient pain, tissue hyalinization and undermining resorption; therefore, the appliance design that is used to apply such force should behave elastically over a period of weeks to months **(Profitt *et al.*, 2013).**

Until the 1930s, gold orthodontic wires are the only type available for clinical use, then further materials with desirable characteristics have been adopted in orthodontics such as stainless steel, nickel titanium, cobalt chromium and others **(Singh, 2016).**

Advancements in material science and technology along with developments in the properties of existing ones have presented some newer arch wire materials. Accurate selection and understanding of the biomechanical requirement of each phase of treatment necessitates accurate characterization researches on arch wire alloys **(Krishnan and Kumar, 2004a).**

The selection of proper grade of wire would provide the benefit of ideal and anticipated treatment outcomes. The clinician must therefore be familiar with the difference in the mechanical properties and clinical application of the various grades of orthodontic arch wires **(Hazel *et al.*,1984; Kumar,1989).**

The demand for the aesthetic modalities is growing among patients seeking orthodontic treatment, the development of the orthodontic arch wires with optimum aesthetic appearance and clinical performance has become an essential and important factor of the treatment nowadays ***(*Huang *et al.*, 2003; Elayyan *et al.*, 2010).**

Different types of aesthetic orthodontic arch wires have been introduced in the market such as Teflon coated arch wires, epoxy coated wires, orthodontic wires coated with a nylon-based matrix reinforced with silicone fibers, and orthodontic wires made from composite material reinforced with glass fiber **(Cardosa and Helena, 2009).**

The development of orthodontic wire technology and new orthodontic techniques have led to further researches for better quality and more biocompatible alloy **(Agwarwal, 2011).**

1.1.2 Types of arch wires

Birnie and Harradine (2005) classified arch wires according to the following characteristics which are:

- **materials:** (stainless steel, titanium, Elgiloy, glass or polymers).
- **Coated or non-coated*:*** (ion implantation or spray coating).
- **Morphology*:*** (round, rectangular, single, multi-strand or braided).

1.1.2.1 Stainless steel arch wires

An iron–carbon alloy composed of 18% chromium and 8% nickel. The most important properties of this alloy is high corrosion resistance which is contributed to the formation of the passivated oxide layer blocking oxygen diffusion to the underlying layers. Chromium and nickel both reduce the critical temperature (temperature at which austenitic structure break down on cooling) so that single phase structure of austenite was stabilized and the overall corrosion resistance was enhanced **(Brantley and Eliades, 2001; Noort, 2002).**

Stainless steel arch wires demonstrate high strength the stored energy of activated stainless steel wires is lower than of beta-titanium or nickel titanium wires with high modulus of elasticity **(Goldberg *et al.*, 1977; Drake *et al.*, 1982; Kusy, 1983).** lower spring-back of than that of the other two types of wires **(Ingram *et al.*, 1986)**, Ductility of stainless steel is adequate **(McCabe, 1985)** and characterized by lower bracket-wire friction **(Kusy and Whitley, 1990**), moderate cost and good formability (**Krishnan and Kumar, 2004a)** and high stiffness and corrosion resistance **(Bock *et al.*, 2008).**

The regular grades of orthodontic stainless steel wire can be bend to almost any desired shape without fracturing. While the super grade wires do not behave similarly; therefore, if sharp bends are not needed, the super wires can be useful, but it is difficult to show improved clinical performance due to high cost and limited formability **(Proffit *et al.*, 2013).**

The properties of stainless steel wires can be controlled over a wide range by altering the amount of cold working and annealing during manufacturing. They can be softened by annealing and hardened by cold working **(Proffit *et al.*, 2013).** Also, these wires can be soldered and welded to make complex appliances, although it is necessary to use solder to reinforce the weld joints **(Pattabiraman *et al.*, 2014).**

According to American Iron and Steel Institute, stainless steel arch wires are categorized into: martensitic, ferrite, austenitic and duplex steel (**Yoo *et al.*, 2008).**

1.1.2.2 Cobalt Chromium wires (Co-Cr)

Co-Cr alloys are available commercially as Elgiloy which was patented by the Elgin National Company **(Kusy, 2002).** This alloy contained 8 elements which are cobalt 40 %, chromium 20 %, nickel 15 %, iron 16 %, molybdenum 7 %, manganese 2 %, carbon 0.16 % and beryllium 0.04 %. These wires are supplied with different physical properties **(Ireland and McDonald, 2003; Anusavice *et al.*, 2012).**

Elgiloy have very good formability prior to heat treatment, which takes place once they are configured, thus increasing their stored energy and functionality **(Kusy, 1997).** They can be heat hardened at 482°C for about 7 minutes after manipulation to increase strength approximately equal to that of stainless steel **(Singh, 2007)**.

Non-heat treated Co-Cr wires have a lesser spring back than stainless steel wires of the same section **(Ingram *et al.*, 1986)** The advantages of Co-Cr wire over stainless steel include greater resistance to fatigue and distortion and easily soldered **(Kapila and Sachdeva, 1989).** Moreover, it has excellent corrosion resistance property **(Reclaru *et al.*, 2005).**

At the end of the recent century, this type of alloy had almost hidden from the market due to its high cost as compared to stainless steel wires and the extra step of heat treatment to obtain the required properties **(Mistakidis *et al.*, 2011; Anusavice *et al.*, 2012; Proffit *et al.*, 2013).**

1.1.2.3 Titanium wires

Titanium alloys are generally used as biomaterials, mainly as orthodontic wires, because of their excellent elasticity and shape memory which allow these materials to return to their original shape after stress relief **(Wichelhaus *et al.*, 2010; Catauro *et al.*, 2014).** There are several types of titanium alloys used in orthodontics, they are:

A-Nickel-Titanium (NiTi) Arch Wires

NiTi alloys have been developed by the Naval Ordinance Laboratories in the USA and marketed as (Nitinol). The growing importance of these alloys depend on the extra ordinary mechanical properties **(Singh, 2007; Bezrouk *et al.*, 2014).**

The phase transition of NiTi alloys provides two main properties which are: shape memory and super-elasticity. Shape memory implies the ability of the wire to return to its previous form after distortion while super-elasticity means the very large reversible strain that certain NiTi wire can tolerate. Therefore, it delivers a low constant force during loading and unloading **(Fernandes *et al.*, 2013; Proffit *et al.*, 2013; Sivaraj, 2013; Takamizawa and Miyamoto, 2014).**

NiTi arch wires are used during leveling and alignment stage of treatment, since they have low modulus of elasticity and high elastic range when compared to stainless steel wires, so they deliver light continuous force over a wide range of activations which is preferred in orthodontic treatment **(Proffit *et al.*, 2013; Senkutvan *et al.*, 2014).**

The additional advantage of NiTi arch wires is their high corrosion resistance and good biocompatibility because of the passivated stable oxide layer that is formed on the titanium surface **(Chen *et al.*, 2013; dos Santos *et al.*, 2014; Katić and Metikoš-Huković, 2014; Nalbantgil et al., 2016).**

Among disadvantages of NiTi, a low release of nickel or titanium may be occurred and nickel hypersensitivity reactions have been observed in orthodontic patients with nickel sensitivity, although these cases are rare **(Heakal and Awad, 2011; Chakravarthi *et al.*, 2012; Maheshwari *et al.*, 2015)** another disadvantage is the load deflection rate decrease with observable surface change in the presence of other materials such as fluoride agents (**Alkhatieeb, 2006; Hashim and Al-Joubori, 2017).**

B- Beta-titanium wires (β-Ti)

β-Ti wires are also known as titanium-molybdenum alloy (TMA) which are introduced in 1979 as an orthodontic wire **(Burstone and Goldberg, 1980)** popularized in orthodontic applications at the present decade **(Proffit *et al.*, 2013; Insabralde *et al.*, 2014).** TMA composed of approximately 11.3% molybdenum, 6.6% zirconium and 4.3% tin, with the balance of titanium (approximately 77.8%). Titanium-molybdenum wires demonstrate good formability, but should not be strongly bent because of a risk of breaking **(Burstone and Goldberg, 1980)**.

Additionally, they have a modulus of elasticity lower than of stainless steel wires and almost twice that of NiTi, and high ductility **(Juvvadi *et al.*, 2010).** The joinability of TMA wires are better than stainless steel wires since they demonstrate higher resilience and better surface and structural characteristics, which indicates only a minor change in wire properties after welding **(Krishnan and Kumar, 2004b; Pattabiraman *et al.*, 2014).**

C- Titanium- Niobium

Titanium-Niobium alloy was introduced by **Dalstra *et al.* (2000)** as a finishing arch wire containing 45% niobium as a stabilizing element. This alloy is nickel free arch wire alloy with a modulus of elasticity nearly (50%) of stainless steel and is similar to that of TMA. Also, it can be welded **(Vijayalakshmi *et al.*, 2009; Graber *et al.*, 2012; Arciniegas *et al.*, 2013; Rerhrhaye *et al.*, 2013).** It has been shown to be less susceptible to fluoride enhanced corrosion **(Huang, 2005)**.

Additionally, this alloy is useful when a highly formable wire with low forces in small activations is required **(Graber *et al.*, 2012).**

1.1.2.4 Esthetic arch wires:

The first attempt to make esthetic arch wires was to camouflage them by a plastic layer coverings. Rocky Mountain Teflon coated SS wires were introduced in the 1970s. Although the appearance was greatly improved, experience with the Teflon coated wires showed that the coating tends to stain and split during use and thus revealing the underlying metal The first non-metalic aesthetic arch wires were introduced in 1992 and manufactured by Oromco **(Postlethwaite, 1992)**.

Aesthetic arch wires **(Fig 1.1)** can be classified into two main categories **(Burstone *et al.*, 2011; Aksakalli and Malkoc, 2013; Mitchell, 2013)** which are:

Figure 1.1 Aesthetic orthodontics (**Aksakalli and Malkoc, 2013)**

1- Metallic coated arch wires

They include NiTi or stainless steel arch wires coated with a layer of polymers or inorganic materials like epoxy-resin, polytetrafluoroethylene (Teflon), synthetic fluoride resins, parylene-polymer or less commonly palladium layer to produce aesthetic wires which mimic the shade of teeth **(Jabbari *et al.*, 2012; Kaphoor and Sundareswaran, 2012; Arango *et al.* 2013; Shashwath *et al.*, 2013).**

Manufacturers differ in the material and thickness of coating and the procedure of application to increase aesthetics and mechanical efficiency **(Zegan *et al.*, 2012; Aksakalli and Malkoc, 2013).**

Rongo (2014) classified metallic coated wires according to the procedure of coating and the material used for coating.

A- According to the procedure of coating

There are different methods to cover the wire with coating materials to improve the surface features **(Santiago *et al.*, 2013).** Previously **Peláez-Vargas (2005)** stated that the coating methods include either chemical or thermal technique.

(1) Thermal methods

(a) Thermal Plasma Spray

This method involves the use of high temperature which lead to melting of any material and depends on the radiofrequency inductively coupled plasma discharge. this is the method of deposition of finely ground materials on a molten or semi-molten metal producing the rough coated layer that is appropriate for orthopedic applications and can be used in dental implantology **(Wang *et al.*, 2009; Junker *et al.*, 2010).**

(b) Chemical vapor deposition

This method has wide applications in dentistry such as coating of endodontic files and burs which involves chemical reactions that occur around the hot surfaces which are placed in a chamber where a precursor gas is flowing resulting in deposition of a thin coating layer on the metal surfaces. This process has not been used in orthodontics yet, but it is under research **(de Vasconcellos *et al.*, 2013).**

(c) Physical vapor deposition

This method has been widely used by many researchers who reported that it affects both the physical and mechanical properties. This process includes the vaporization of the material and using the vapor in its atomic level after its being transformed to a very large number of nanometers for freestanding structures, multilayer coverings, graded composition deposits and very thick deposits **(Mattox, 2010; Krishnan *et al.*, 2011; Ryu *et al.*, 2012).**

(2) Chemical Methods

(a) Electrodeposition

In this method **(Fig 1.2)**, the cathode is represented by the substrate to be coated whereas anode is represented by the metal to be deposited in form of salts dissolved in water maintaining a controlled temperature. This technique decreases the friction of coated arch wires as it is reported by many studies (**Redlich *et al.* 2008; Samorodnitzky-Naveh *et al.* 2009).**

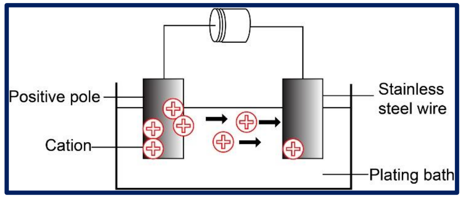

Figure 1.2 Electrodeposition (Usui et al., 2017)

(b) Sol-Gel Method (Fig 1.3)

The temperatures used for the production of ceramic, glass and ceramic-glass materials are less than that of the other techniques, different shapes can be obtained such as nanospheres or monolith **(Patil *et al.*, 2004; Peláez-Vargas, 2005; Arango *et al.*, 2013).** This technique is used to coat stainless steel orthodontic arch wires and it is useful to reduce or eliminate friction between wire and bracket and inhibit plaque growth on the appliance during treatment **(Rendón *et al.*, 2008).**

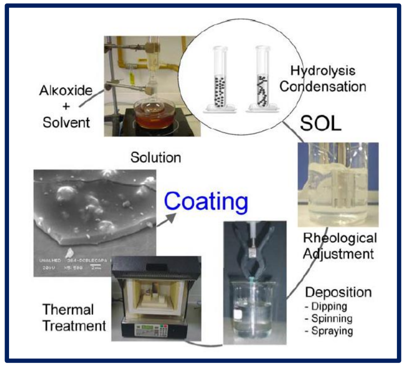

Figure 1.3 Sol-Gel method **(Arango *et al.*, 2013)**

B- According to the material of coating

(a) Teflon coated wires

Polytetraphloroethylene (PTFE), currently known by the trade name "Teflon", is an aesthetic and anti-adhesive synthetic polymer which consist mainly of fluorine and carbon. The fluorine carbon bonds make Teflon layer hydrophobic in nature, and heat resistant. It gives a plastic tooth colored coating so that it can blend with the teeth and aesthetic brackets colors **(Farronato *et al.*, 2012; Kravitz, 2013; Goyal, 2014; Rongo, 2014; Singh, 2016).**

Teflon coating is produced by a thermal coating technique **(Burstone *et al.*, 2011; Rongo, 2014)**. These coated wires have less friction as compared to uncoated wires **(Husmann *et al.*, 2002; Farronato *et al.*, 2012; Rongo, 2014).**

(b) Epoxy coated wires

Epoxy is a synthetic resin that consist of epoxide with another compound. It is applied to arch wires by electrostatic technique in which electrostatically charged elements are used for coating **(Bandeira *et al.*, 2011; Kravitz, 2013; Nascimento *et al.*, 2013; Rongo, 2014)**.

Epoxy coated wires have lower frictional forces as compared with uncoated arch wires **(Elayyan *et al.*, 2010; Bandeira *et al.*, 2011; Alavi and Hosseini, 2012).**

(c) Parylene-polymer

A new coated material has been developed with silver covering to provide a thin aesthetic coating**.** The surface of parylene coating material has rougher morphology and low hardness but they have similar mechanical properties as compared with the uncoated arch wires **(Lijima *et al.*, 2011).**

❖ Advantages of coated arch wires

- Coated arch wires provide the desired aesthetics corresponding with the aesthetic brackets, increased electrical resistance, high toughness and good thermal conductivity **(Wichelhaus *et al.*, 2005; Piel *et al.*, 2011).**
- Protect the underling metal wire from corrosion **(Neumann *et al.*, 2002; Rongo, 2014).**
- The coatings reduce friction during treatment which is necessary in the beginning stages of treatment, whereas a large coefficient of friction is essential in case of anchorage **(Husmann *et al.*, 2002; Burrow, 2009; Bandeira *et al.*, 2011; Farronto *et al.*, 2012; Krishnan *et al.*, 2013; Bravo *et al.*, 2014).**

❖ Disadvantages of coated arch wires

- Several studies concluded that coated wires deliver lower forces when compared to uncoated wires of the same nominal diameter **(Elayyan *et al.*, 2010; Alavi and Hosseini; 2012; Bradley *et al.*, 2014; Pop *et al.*, 2015).**
- It was concluded that 25% of the coating material is removed in 33 days intraorally. Consequently, the underlying metal wire is exposed **(Quintao and Brunharo, 2009; Bradley *et al.*, 2014).**
- Coating layer may alter the thermal properties of wires as it works as a separating layer for heat transfer **(Bradley *et al.*, 2014).**

2- Non-metallic aesthetic arch wire (Transparent)

Within the last decade of recent century, significant attempts have been made to produce nonmetallic arch wires with properties similar to metallic alloys. Dr. Talass in 1992 design the first non-metallic wire which was named Optiflex arch wire (Figure 1.4). Optiflex is the most esthetic arch wire made of clear optical fiber that consist of 3 layers **(Agwarwal *et al.*, 2011; Shashwath *et al.*, 2013; Kotha *et al.*, 2014; Rongo, 2014)**:

- the inner core is silicon dioxide core which provide the force for teeth movement.
- the middle layer is silicon resin which protect the core from the moisture and provides some strength to underlying layer.
- the outer layer is nylon layer which prevent the damage to the wire and provide further strength.

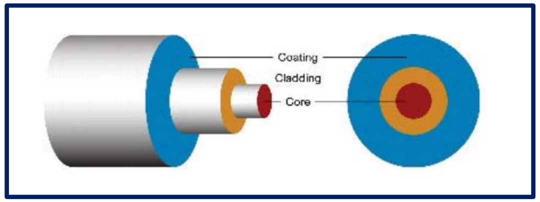

Figure 1.4: Layers of Optiflex wire (Singh, 2016)

Lim et al. (1994) refuse Optiflex arch wire due to its demonstrated low force and stiffness and poor spring-back as compared to regular wires.

Nowadays**,** the two main types of transparent non-metallic wires are the non-formable fiber-reinforced and the formable self-reinforced composite alternatives which are desirable for the leveling and alignment phases in Class I malocclusion **(Proffit *et al.*, 2013; Valeri, 2013; Rongo, 2014)**.

- **Fiber-Reinforced Polymer Composite (FRPC) arch wires**

FRPCs have been used in several dental applications, such as: endodontic posts and cores, fixed partial dentures, space maintainers, periodontal and orthodontic splints, and recently for implants **(Kathuria *et al.*, 2011; Kumbuloglu *et al.*, 2011; Nidhi *et al.*, 2012; Alavi and Mamavi, 2014; Chng *et al.*, 2014a*)***.

In 2008, Biomers® received approval from food and drug administration agency for clinical use of FRPC archwire, which is formerly marketed as the first totally clear archwire. They are presented as round and rectangular cross-sections, and can be paired with esthetic pre-torqued and pre-angulated brackets according to the practitioner desire **(Proffit *et al.*, 2013; Shashwath *et al.*, 2013; Woods, 2013; Chng *et al.*, 2014b)**.

The fibers used for strengthening may be short or continuous filaments. Short fibers are arranged parallel to the long axis of the wire and resulting in a low stiffness wire. Continuous fibers are aligned parallel to each other along the long axis of the wire which lead in a large range of spring-back and elastic recovery ***(Valiathan and Dhar, 2006; da Silva et al., 2013; Tanimoto et al., 2015)***.

The main advantage of FRPC arch wires is their appearance, because of their good transmission capacity of the color of teeth, so they are improving aesthetics with the use of ceramic brackets (***Chng et al., 2014a; Chng et al., 2014b; Inami et al., 2016)***. FRPC arch wires are susceptible to fracture when activated more than 3 mm which is considered as a main disadvantage of these wires; Furthermore, water and fluoride immersion can affect the mechanical properties and causing damage to the wire surface **(Burstone *et al.*, 2011; Antonopoulou *et al.*, 2012; Doshi and Mahindra, 2013; Ohtonen *et al.*, 2013; Chang *et al.*, 2014; Alobeid *et al.*, 2017).**

- **Self-Reinforced Polymer Composite (SRPC) arch wires**

Newly introduced SRPC wires are fiber-free composed of polyphenylene polymers. They are not currently available but in the development phase close to be introduced to orthodontic market as round and rectangular cross-sections arch wires **(Proffit *et al.*, 2013; Valeri, 2013; Bradley *et al.*, 2014)**.

Mechanical properties are comparable to NiTi and β-Ti wires with somewhat smaller cross-sections. Strength and stresses delivered are slightly lower than typical, but their values are adequate for the first stage of the orthodontic therapy (**Kuhlberg, 2009; Burstone *et al.*, 2011).**

1.1.3 Requirement of an ideal arch wires

The properties of an ideal arch wire can be described in the following principles, but in contemporary practice, no arch wire meets all these requirements, and the best outcome can be achieved by using specific arch wire for a specific purpose **(Kusy, 1997)**, the requirements include:

- High resistance to corrosion caused by fluids **(Graber *et al.*, 2017).**
- The arch wires should be efficiently ductile to resist breakage due to sudden loading in the mouth **(Graber *et al.*, 2017).**
- The arch wire should be able to be designed in a soft state and later treated by heat to become harder **(Graber *et al.*, 2017).**
- The arch wire alloy should permit soldering of attachment easily **(Graber *et al.*, 2017)**.
- Arch wires should be biocompatible with poor biohostability **(Kusy, 1997).**
- Wires should have low coefficients of friction, ideally little or no friction should be between the arch wire and bracket **(Kusy and Whitley, 2000)**. Friction between arch wire and bracket slot can compromise the efficiency of tooth movement **(Iwasaki *et al.*, 2003).**
- Low cost **(Jones and Newcombe, 1995).**

- The arch wire should deliver light constant force to motivate osseous development **(Waters, 1981; Gurgel *et al.*, 2001)**; and it should be elastic for weeks or months **(Gurgel *et al.*, 2001; Ogawa *et al.*, 2002).**

1.1.4 Mechanical properties of arch wire materials

The mechanical properties of orthodontic arch wires are very important and should be well understand in order to achieve a clinical success **(Vijayalakshmi *et al.*, 2009).** The mechanical characteristics of a material are determined by numerous factors and the intrinsic properties are determined by material composition at the molecular level while extrinsic properties are macroscopic features of the material, such as wire dimensions **(Nanda, 1997).**

To compare between the performance of any materials in different applications, we need standardizations of comparison that do not depend on the size and form of the material. This enable to predict the behavior of the elements that the material made from them. The basis of such standardization are the quantities which can be named stress and strain **(Noort, 2002).**

Stress is defined as the internal resistance of a material to the applied force and depend on the direction of this force; which are bending, tensile, compressive, shear, and torsion **(Craig and Marcus, 1997).**

Proffit (2013) defined stress as the force per unit area while strain as a deflection per unit length. And the unit of stress measurement (the standard national unit) is a newton per meter squared (N/m2), whereas the strain is dimensionless.

A conventional stress-strain curve is important since it provides a way for obtaining data about materials' tensile or compressive strength **(Hibbeler, 2003).**

The load deflection curve (Figure1.5) describe the load or force exert on a material to the distortion (deflection) and both areas of that curve can be described **(Nanda, 1997; Proffit *et al.*, 2013; Graber *et al*., 2017)** as follow:

A. Elastic Region: It is the linear portion of the curve. Deformation of the material in this area is temporary that the material will return to its original form with force relief.

B. Plastic Region: It is the area of the curve where the material is permanently deformed.

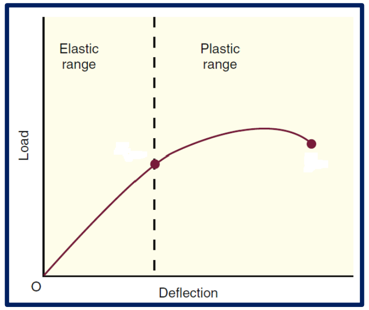

Figure 1.5 Typical load-deflection curve (Graber, 2017).

There are several basic mechanical properties **(Figure 1.6)** of orthodontic arch wires that can be determined from the laboratory tests (**Proffit *et al.*, 2013)** which are**:**

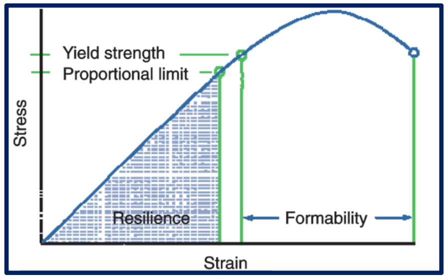

Figure 1.6 Mechanical properties from stress strain curve (Proffit *et al.* 2013).

A. Yield Strength: It is a convenient measure of the proportional limit or the point beyond which permanent deformation will happen with increased force **(Drake *et al.*, 1982)** and represents the stress below which deformation is completely elastic. It indicates the amount of energy stored in an orthodontic wire before it is plastically deformed **(Kesling, 1987).**

B. Stiffness: It is a main determinant of the force applied to the teeth and a measure of the force required to bend a wire to certain distance. The larger the force, the farther it will bend. The stiffer the wire, the more force it taken to bend it by which more force delivery will be happened. (**Creekmore, 1976; Proffit *et al*., 2013).** stiffness is affected by:

- **Wire manufacturer process:** including formula of the alloy, heat treatment and hardness **(Creekmore, 1976).**
- **Wire shape and size:** for round wires, stiffness is proportional to the fourth power of diameter which mean a 0.022-inch wire needs 16 times as much force to deflect it the same distance as an 0.011-inch wire. A force of one pound is essential to bend a 0.022 inch wire the same distance as one ounce of force will bend an 0.011-inch wire. So smaller round arch wires will deliver lighter forces during alignment. While with rectangular wire, stiffness is directly proportionate to width and proportionate to the cube of thickness. Doubling the width doubles stiffness. Doubling thickness increases stiffness by 8 times **(Garrec *et al*., 2005).**
- **Length of span:** length of span affects the function of all beam types in exactly the same way but with different standard, so in Bending test, stiffness is inversely proportional to the cube of length if it requires a half a pound of force to deflect a wire a certain distance, one ounce of force will deflect the same wire the same distance if the length of span is doubled. While in Torsion, stiffness is inversely proportional to length. Doubling the length will cause the ends to twist twice **(Creekmore, 1976).**

- **Temperature:** it could affect the bending stiffness of super-elastic NiTi arch wires with change of mouth temperature **(Meling and Odegaard, 2001).**

Modulus of elasticity is the measure of stiffness of material and can describe the resistance of materials to be deformed which is considered as an important clinical factors that closely affect the biological nature of teeth movement. Low modulus means low stiffness which is needed in the beginning stages of treatment to minimize patient discomfort and improve biological tooth movement; while, in the finishing stage of wires to contain the movement of teeth achieved earlier, a stiffer (high modulus) wire is required **(Burstone, 1981, proffit *et al*., 2013)**.

C. The maximum spring-back (range): It is also referred to as maximum elasticity, range of deflection or range of activation or working range **(Ingram *et al*., 1986).** And can be defined as the distance that the wire will elastically bend before permanent deformation and measured in length unit as in millimeters (**Barrowes, 1989**).

D. Resilience: the ability of the wire to store energy and represent a combination of springiness and strength **(Profit, 2013).**

E. Formability: It is defined as the property that allow the practitioner to form the wire into the required design with certain configuration such as coils, stops and loops without breakage **(Burstone and Goldberg, 1980)**. It represents the amount of permanent deformation that the wire will withstand before failure or the permanent bending that the wire will tolerate before it breaks **(Proffit, 2013).**

F. Ultimate Tensile Strength: it is the maximum force that the wire can withstand before the material begins to weaken; and represents the maximum point of the force deflection curve **(Britto and Isaacson, 2001).**

G. Welding and soldering: the biomechanical attachments are important in orthodontic wires and can be achieved by electric welding or by using a bonding agent. Welding of attachments to the arch wire is achieved for appropriate force delivery and can be done when the arch wire has superior weld features **(Krishnan and Kumar, 2004b).**

H. The proportional Limit: is the maximum stress that a material will tolerate without a deviation from the proportionally of stress to strain, below the proportional limit, no permanent deformation occurs in a structure **(McCabe, 1985).**

I. The Elastic Limit: is defined as the stress just before permanent deformation takes place **(Burstone, 1981),** and it is the maximum stress that the material can withstand before permanent deformation **(Waters *et al*., 1981).**

There are several methods that can be used for measurements of mechanical properties of arch wires which are:

1- Tensile test (Tension test)

The wire is usually fixed at one end and then stretched longitudinally, the results of tensile test is too complex to be easily understand, tensile forces are not usually applied in clinical orthodontics **(Yang *et al.* 2001).**

2- Cantilever Configuration:

The mechanical properties of wires can be typically determined under bending conditions, which considered more representative to the type of deformation that occurred clinically than the tensile test **(Asgharnia and Brantley, 1986)**.

In cantilever bending, the forces are assessed as bending moments, and are expressed in grams per millimeter (gm/mm). The deflection is measured in degree **(Brantley and Myers, 1979).**

3- Three-point bending test.

The three-point bend test (Figure 1.7) is of choice in orthodontic literature for measurement of the stress, strain, strength and modulus of elasticity of orthodontic arch wires. Several researches have been confirmed using this test to estimate the load-deflection properties, which in turn can be used to determine and predict the biological nature of tooth movement (**Krishnan and Kumar, 2004a).**

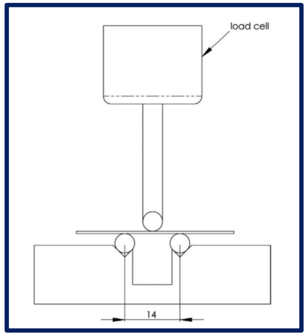

Figure 1.7 Three-point bending test (Bartzela, 2007).

When a wire is subjected to a bending force, the internal surface is compressed, whilst the outside surface is subjected to tension. The bending stress

is equal to zero at the neutral axis, which is the middle area between the two surfaces. The neutral axis resembles a flat ribbon which passes through the center of the wire between the two curved surfaces dividing that stress into tensile and compressive one **(Cacciafesta *et al.*, 2008).** The maximum stress is developed at the surface of the wire, directly opposite the applied load and it gradually decreases to zero at the supported ends **(Mencik, 1992).**

1.1.5 Arch wires and ideal force production

Orthodontic arch wires deliver biomechanical forces required to move or stabilize the teeth. Orthodontists are always looking for the most effective and efficient arch wire. A perfect arch wire should move the teeth with light continuous forces. Generally, orthodontic forces are ranged from 1.5-5 N **(Holberg *et al.*, 2008). Proffit *et al.* (2013)** reported the optimum forces required for orthodontic movements, as shown in table 1.1. Therefore, knowledge of the biomechanical behavior and clinical applications of orthodontic wires is essential for accurate application of the treatment plan.

Table 1.1: Optimum Forces for Orthodontic Tooth Movement (Proffit et al., 2013).

Type of Movement	Force (gm)
Tipping	35 – 60
Bodily movement	70 – 120
Root up righting	50 – 100
Rotation	35 – 60
Extrusion	35 – 60
Intrusion	10 – 20

1.2 Friction

Kusy (2005) defined friction as the resistance to the movement of two or more contacting bodies. Several factors influence the friction and it is very difficult to isolate individual factors **(Wichelhaus *et al.*, 2005)**.

Or friction, defined as the force that resist movement, in which two surfaces slide over each other, and has a multifactorial nature **(Braga *et al.*, 2011).**

Friction resistance is a frequently used term that refer to the force resisting the sliding movement of the teeth. Whereas, resistance to sliding is more proper term, as it can be classified into three phenomena: classic friction, binding, and notching **(Alsubie and Talik, 2016).**

In orthodontics, friction compete with tooth movement whenever sliding mechanics are involved. During sliding mechanics, the wire contacts the bracket and ligation and a frictional force occurs in the opposite direction and against the orthodontic force thus decreasing its magnitude **(Vandeberg, 2008).**

1.2.1 Laws of Friction

Coulomb's model of friction **($F = N \times \mu$)** states that the maximum frictional force is proportional to the magnitude of the normal contact force **(N)** multiplied by **(μ)**, the latter representing a coefficient of friction. The normal force **(N)** is the force that is perpendicular to the shared area of contact between the two objects **(Blau, 1996)**.

1.2.2 Factors Affecting Friction

1- Physical factors

A. Arch wire dimension

As a wire dimension increases, the contact between the wire and the bracket slot increases. Several studies have proved that large wires produce more friction during sliding through the brackets. It has been shown that smaller wires

produce less friction because of the greater free space in the slot and their larger elasticity **(Drescher *et al.*, 1989; Kusy and Whitley, 1997).**

B. Arch wire Shape

Arch wires come in two typical forms; round and rectangular. Numerous studies show that rectangular wires produce more friction than round wires but only in certain conditions **(Drescher *et al.*, 1989; Ogata *et al.*, 1996; Mendes and Rossouw, 2003**).

Drescher *et al.* (1989) found that the occluso-gingival dimension of the wire was the most critical dimension affecting friction.

C. Arch wire material

Stainless steel arch wires have generally been the most widely used wires in orthodontics and it has been found that stainless steel wires have a lower bracket-wire friction than other types of wires (**Krishnan and Kumar, 2004a).** Other common alloys have been developed in the last decades because they have good properties such as Elgiloy, NiTi, and TMA arch wires **(Kusy and Whitley, 1997).**

These types of arch wires have varying degree of friction since they are made from different materials. Stainless steel arch wires produce less amount of friction with stainless steel brackets, while Elgiloy and NiTi wires produce more friction than stainless steel but in similar amounts, while TMA produces the highest amount of friction **(Frank and Nikolai, 1980; Drescher *et al.*, 1989; Kusy and Whitley, 1999a, Jian-Hong, 2011).**

D. Arch wire surface texture

The basic rules of friction state that coefficient of friction do not depend on the area between the two contacted objects; Although, there are several studied have proved that roughness is directly proportional with friction. (**Frank and Nikolai, 1980; Kusy and Whitley, 1997**). It has been shown via laser spectroscopy that different arch wire alloys have significantly different surface textures. Stainless steel appears the smoothest, followed by Elgiloy, TMA, and NiTi. **(Kusy and Whitley, 1988);** however, **Doshi (2011)** concluded that there is no relationship between surface roughness and friction.

E. Stiffness of arch wire

The stiffness of an arch wire is determined by the modulus of elasticity, the resistance to sliding depend on wire stiffness **(Rucker and Kusy, 2002)**. For a given arch wire size, a stainless-steel wire is stiffer because of its larger modulus of elasticity value than NiTi wire; thus the stainless-steel wire cannot negotiate the edges of the bracket as done by NiTi wire **(Thorstenson and Kusy, 2002)**.

A sufficient clearance should be present between the bracket slot and the wire to avoid binding, notching and resultant increased friction (Fig 1.8). The clearance or play in the second order depends on many factors such as bracket slot size and arch wire dimension **(Nanda, 1997)**.

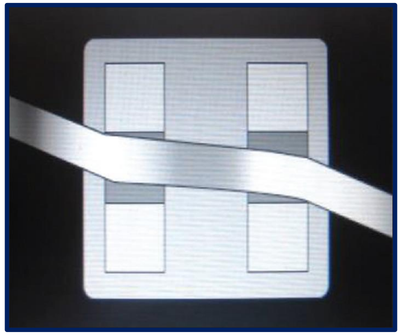

Figure 1.8 Binding and notching of arch wire (Prashant *et al*., 2015)

F. Esthetic arch wires

Esthetic arch wires are the greatly desired types to complement esthetic brackets in clinical orthodontics, so several types of aesthetic wires have been developed for this reason **(Brantley and Eliades, 2001).**

The frictional characteristics of a polymeric esthetic arch wire show a binding during sliding with increased friction between the wire and bracket slot; moreover, a plastic deformation has been proved even with low forces **(Omana *et al.*, 1992).**

A totally esthetic labial arch wire was presented to the market in 1991, It was made of clear optical silicon fiber, and has been considered a perfect wire for initial leveling and alignment with little friction **(Kusy, 1991).**

Tooth colored plastic coated NiTi wires are also available. These types of wires have a stain and crack resistant coating layer followed by a silicone coating to reduce friction. The wire is fabricated from NiTi wire that is 0.002 inches thinner to compensate for the thickness of coating layer. Ion implanted titanium wires have recently become available. This offers an esthetic alternative to the regular TMA **(Mendes and Rossouw, 2003).**

The frictional force between the bracket slot and arch wire is a primary subject in orthodontics because it restricts dental movement **(Burrow, 2009)**. Some studies **(Choi *et al*., 2015; Rudge, 2015)** have investigated that frictional properties of aesthetic arch wires are primarily focused on the relationship with surface roughness.

The ion implantation with rhodium coating and the Teflon coating are the most Popular surface treatment for aesthetic arch wires. Rhodium coated wires show the greatest amount of surface roughness with increased coefficient of friction **(Husmann *et al*., 2002; Elayyan *et al*., 2010).** While aesthetic coated arch wires with Teflon show the least roughness among other types of coated wires and this improve sliding of wire through brackets slots **(Husmann *et al*.,**

2002; Neumann *et al.*, 2002; Wichelhaus *et al.*, 2005). Consequently, Teflon Coated arch wires could be a possible way to decrease friction **(Farronato *et al.*, 2012).**

G. Ligation

Stainless steel ligature wires were commonly used in the late century until the introduction of elastomeric ligatures, ligation is possibly produce some friction with the arch wire thereby restricting orthodontic tooth movements **(Vande-Berg, 2008).**

Frank and Nikolai (1980) have compared between friction with stainless steel ligatures and elastomeric ligatures and found that the friction increase with increased force applied on the wire with no significant differences between elastomeric ligature and stainless steel ligature tied with a force of 225 grams.

H. Bracket material and design

Although not as esthetically pleasing as plastic or ceramic brackets, the stainless-steel brackets were an esthetic improvement over previously used bands and become most brackets used thereafter (**Proffit *et al.*, 2013).**

Stainless steel brackets have lower frictional forces when compared with ceramic brackets; which may be contributed to the smooth surface of stainless steel brackets **(Kusy and Whitley, 2001; Cha *et al.*, 2007; Williams and Khalaf 2013)**.

Titanium brackets were introduced to be more biocompatible than stainless steel and withstand several conditions in the oral environment. Although the rougher surface of these brackets than stainless steel, they have coefficient of friction similar to that of stainless steel brackets in the passive configuration due to the chemistry of the surface layer that is passivated with a layer of carbon,

oxygen, titanium and nitrogen, similar to the passivated layer of stainless steel brackets which could be the reason of reduced friction **(Smith *et al.*, 2003).**

With the increased demand for aesthetic orthodontic, several types of brackets have been introduced such as Ceramic, polycrystalline alumina, single crystal alumina, and polycarbonate brackets **(Smith *et al.*, 2003).**

Pillai *et al.* (2014) compared the frictional resistance of self-ligating ceramic, composite and stainless steel brackets, and found that composite brackets have less friction resistance while ceramic brackets show maximum frictional force as compared to stainless steel and composite brackets.

To reduce friction of ceramic brackets, some manufacturers made them with a stainless steel slot in order to reduce friction with the arch wire **(Nishio *et al.*, 2004).**

I. Slot size

The two standard occluso-gingival slot heights of brackets are either 0.018ˈ or 0.022ˈ. by decreasing the slot size, the binding between arch wire with the bracket edge is increased due to decreased free space available for the wire which lead to increased frictional resistance force, so the orthodontist should be aware with the use of 0.018ˈ bracket slot dimension **(Kusy and Whitley, 1999a);** However, some studies conclude that there is little effect of slot size and friction **(Tidy, 1989).**

J. Bracket width and inter bracket distance

Several studies demonstrate the effect of bracket width on frictional resistance force, since they are available with different widths. by increasing the mesio-distal width, the friction with the arch wire is increased (**Frank and Nikolai, 1980; Husain and Kumar, 2011).**

The Begg (0.05') and Edgelok (0.12') brackets are frictionless to some extent with low angulations despite the great difference in width and the Lewis anti-tip bracket had the greatest bracket width (0.177') with least amount of friction at high binding angulations (**Frank and Nikolai, 1980).**

K. First - order bend (in-out)

A rotational tendency produced during mesial or distal directed forces that are delivered to the bracket which can be avoided by the first order couple that is established between brackets, wire and ligature for such movements and this causes an elastic binding to the ligature leading to increased frictional resistance **(Kusy and Whitley, 1997; Kusy and Whitley, 1999b).**

L. Second -order bend (Angulations)

An increased angulation between the bracket and the arch wire lead to greater friction during sliding movement **(Ogata *et al.*, 1996)** which could be attributed to the binding of arch wire **(Kusy and Whitly, 1999b; Zufall and Kusy, 2000).**

M. Third - order bend (Torque)

According to physics laws, torque can be defined as a vector that measures the force tendency to rotate an object around some axis (**Serway and Jewett, 2003).** Currently, there is no evidence on the torque with various bracket – arch wire combinations which could be attributed to the difficulty of the experimental configuration required in laboratory studies in addition to other factors including variability of occlusion, differences in individual responses and utilities affecting clinical torque (**Alkire *et al.*, 1997; Harzer *et al.*, 2004).**

During anterior teeth retraction, frictional resistance is increased with third-order torque in posterior segments when the arch wire is sliding through self-

ligating bracket. Frictional forces are extensive, regardless of ligation method if the wire-slot torque exceeds the third order clearance (**Chung *et al.*, 2009)**.

2- Biologic Factors

A lot of in-vitro studies demonstrate friction under dry condition and don't fully simulate oral environment. In the oral cavity, the appliance is pathing in a viscous saliva, plaque and food debris, all of these factors determine amount of friction. To simulate these conditions, some studies use a human or artificial saliva to demonstrate their effect on friction **(Smith *et al.*, 2003).**

Other studies have found that saliva present no significant reduction in frictional forces **(Thorstenson and Kusy, 2001).**

Forces of occlusion is an important factor that play a role in determining the amount of friction in orthodontic appliance and this does not normally found in experimental studies **(Braun *et al.*, 1999).**

Iwasaki *et al.*, (2003) tried to simulate the masticatory function ex-vivo by repeated vertical vibration of an arch wire under different loads (25, 50, 100, 150, 250, and 400 g) with low frequency (91.3 cycles/min) vibration. They found that the resistance to movement through bracket was decreased by this repeated vibration of arch wire. These results agree with other experimental researches and coordinate with them **(Frank and Nikolai, 1980; Drescher *et al.*, 1989; Eliades and Bourauel, 2005).**

Experimental studies often involve laboratory tests with using clean samples, so no plaque or calculus are found to affect friction, the presence of these factors may reduce friction by formation of lubricated environment through salivary proteins adsorption and plaque while the presence of calcified areas may increase friction by increasing surface roughness **(Eliades and Bourauel, 2005)**.

1.2.3 Advantages of friction:

The advantage of friction is relatively restricted in providing anchorage for other teeth movements, however this type of friction may lead to unwanted tooth movements **(Millett and Wilbury, 2004)**.

1.2.4 Disadvantages of friction

An enough amount of force is required to overcome the static frictional forces that is produced between the arch wire and brackets during sliding mechanics, thereby, anchorage loss may be greater than it would be expected if static frictional forces are less considerable **(Jassim, 2006).**

The force that is required for orthodontic tooth movement must be large enough to overcome the friction, this amount of force has a reactive one on the posterior teeth in the mesial direction which is referred to as anchorage loss in clinical orthodontic and should be avoided, so orthodontic materials with low friction values are desired for straight wire mechanics which can reduce anchorage loss **(Nightingale and Sandy, 2001).**

Magnification of the floor of ceramic bracket slot after sliding mechanics show that there was a shiny spot of arch wire debris which is increased as a result of high friction, this can be considered as disadvantage of friction in orthodontic. (**Al-Nasseri, 2000; Silva *et al*., 2010**).

1.2.5 Frictional Control in Orthodontic Appliance

Friction in orthodontic appliance cannot be eliminated from materials, therefore the best key is to control friction by maximizing the efficiency of orthodontic appliance meaning the amount of friction with respect to the applied force and improving the ability of clinician to activate the appliance in a predictable manner; therefore, the clinician should be aware of the characteristic of the appliance that contribute to increased friction and dealing with the force expected to minimize friction **(Rossouw, 2003).**

Many frictionless techniques can be used to simplify teeth movement as in beg technique **(Tidy, 1989).** Many modifications can be made to reduce friction such as using self-ligating brackets which eliminate the forces of ligation and thus reduce friction **(Harradine, 2003).**

Modifications of arch wire can also be made to reduce friction such as increase surface hardness by using ion implanted TMA wires **(Kusy *et al.*, 2004),** or coating the surface of the arch wire by coated materials that contribute to reduce friction such as composite coated arch wires **(Zufall and Kusy,2000).**

Surface coating of orthodontic arch wires with diamond-like carbon have been believed to decrease frictional resistance by incorporating ions on the surface of wire during manufacturing, increasing surface hardness and consequently decreasing friction **(Muguruma *et al.* 2011).**

frictionless elastomeric ligatures have also been prepared from special polyurethane mix to reduce frictional resistance, with an anterior modification that is more rigid and similar to self-ligating bracket cap. These ligatures are recommended when reduced friction is desired **(Thorstenson and Kusy, 2003; Baccetti and Franchi, 2006; Franchi and Baccetti, 2006; Yeh *et al.*, 2007)**. Teflon coated ligature wire have also been made as a method for reduced friction which eliminate the large force produced by stainless steel ligature wire **(Mckamey and Kusy, 1999).**

Chapter Two

Materials & Methods

Chapter Two

Materials and Method

2.1 Materials

2.1.1 The samples (Fig 2.1 and table 2.1)

- Coated stainless steel wires supplied from seven brands ((TP Orthodontic, Rocky mountain (RMO), Hubit, DB Orthodontic, G&H Orthodontic, highland metals, and DANY)) with cross section dimension 0.016'' × 0.022'' and 0.019'' × 0.025'',
- Control group wires (uncoated) supplied from IOS company with cross section dimension 0.016'' × 0.022'' and 0.019'' × 0.025''.

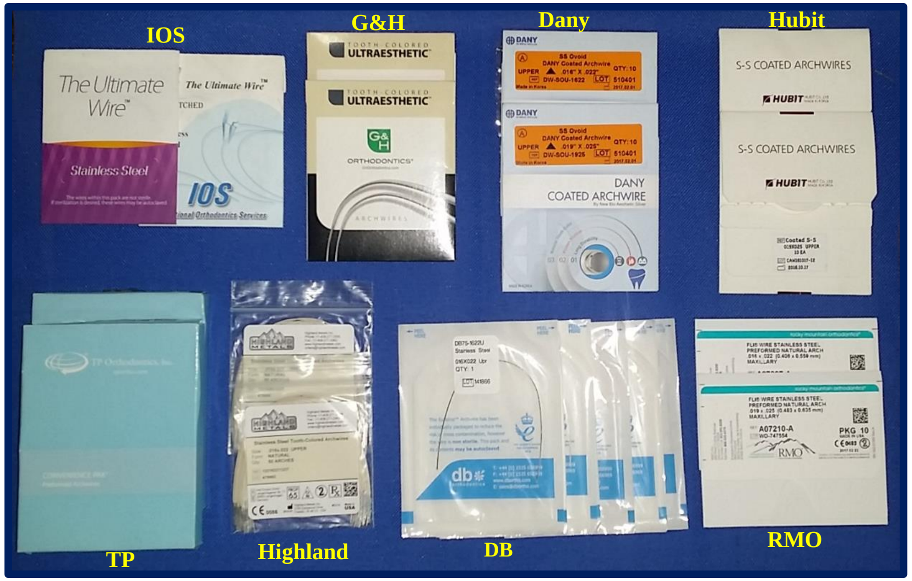

Figure 2.1: The samples (coated and un-coated stainless steel arch wires)

Table 2.1: coated stainless steel arch wires with specific details

Brand name	Trade name	Type of coating	Coated surfaces
DANY BMT (Korea)	Tooth colored arch wire	Polymer	All surfaces
G&H Orthodontics (USA)	Tooth colored arch wire	Epoxy-resin	All surfaces
Highland Metals (USA)	Tooth colored arch wire	Epoxy-resin	All surfaces
HUBIT (Korea)	Tooth colored coated arch wire-perfect	Teflon	All surfaces
DB Orthodontics (UK)	Micro-coated aesthetic arch wires	Teflon	Labial surface
Rocky Mountain (USA)	FLI® Wire	Teflon	Labial surface
TP Orthodontics (USA)	Shiny bright aesthetic wire	Polymer	Labial surface

2.1.2 The Devices

1- Computerized Instron H50KT Tinius Olsen testing machine (England) with a 10 N load cell **(Fig 2.2)**.

Figure 2.2: Instron (Universal testing machine)

2- Metallurgical incident light microscope model Mbi 3300 **(Fig 2.3)**

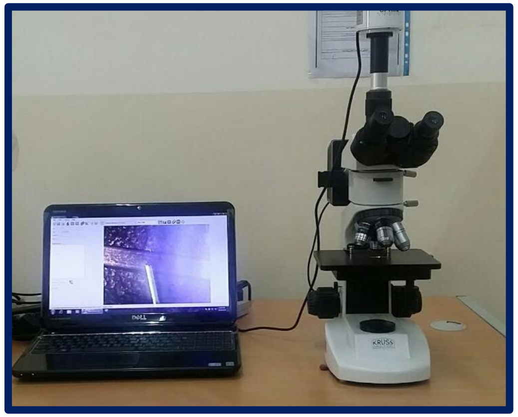

Figure 2.3: Metallurgical microscope (Germany)

2.1.3 Orthodontic materials (Fig 2.4)

1. Ceramic brackets supplied from HUBIT company (Korea)
2. Ligature elastics from Opal company (USA)

Figure 2.4: Orthodontic materials

2.1.4 Orthodontic instruments (Fig 2.5)

1. Orthodontic light wire cutter supplied from Orthotechnology.
2. Bracket holder supplied from Medesy company.
3. Artery forceps.
4. Dental probe.
5. Dental tweezer.

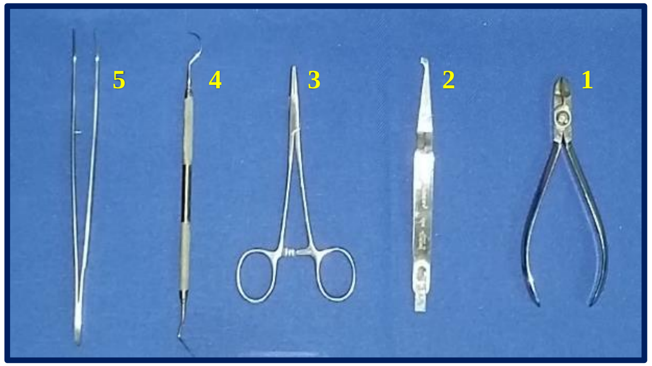

Figure 2.5: Orthodontic instruments

2.1.5 Other materials (Fig 2.6)

1. Gloves.
2. Permanent marker.
3. Ruler.
4. Digital micrometer.
5. Cyanoacrylate adhesive (AMIR).
6. Aluminum blocks.

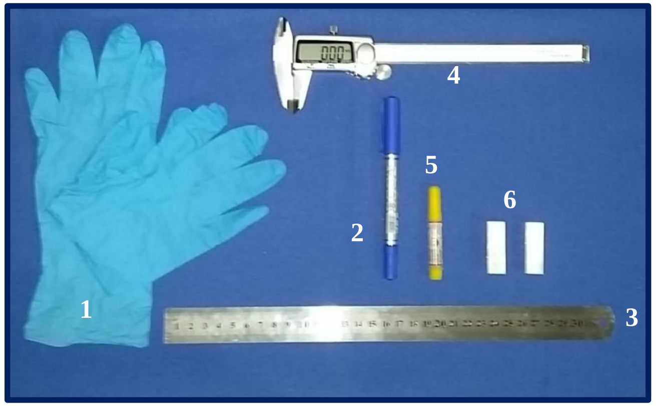

Figure 2.6: Other materials

2.2 The Method

2.2.1 Specimen Selection (Fig 2.7)

238 maxillary coated stainless steel arch wire segments from seven brands (TP Orthodontic, Rocky mountain, HUBIT, DB Orthodontic, G&H Orthodontic, Highland Metals, and DANY Brands) and 34 maxillary non-coated control stainless steel wire segment (IOS) with cross section dimension 0.016'' × 0.022'' and 0.019'' × 0.025'' were used for the three-point bending test, frictional resistance test and for wire dimensions' measurement.

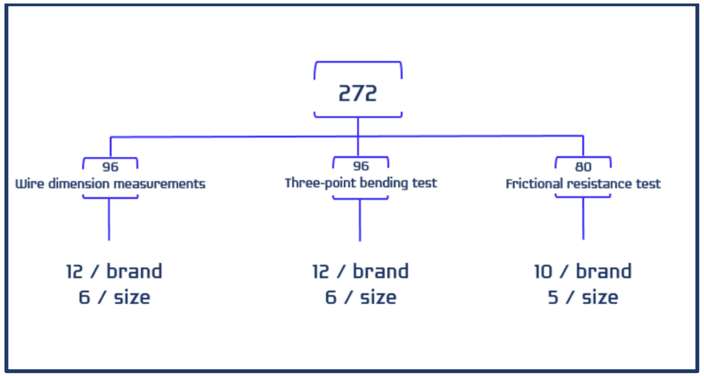

Fig 2.7 Specimens of the study

2.2.2 Preparation of Specimens

For three- point bending test, ninety-six samples with both wire dimensions (twelve for each brand, six for each dimension) were prepared using the following protocol according to ISO standard 15841:2014: A 30 mm wire section should be cut from the straightest part of the pre-curved arch wire.

The preformed arch wires contain a small uncoated part which were cut and discarded **(Fig 2.8)**

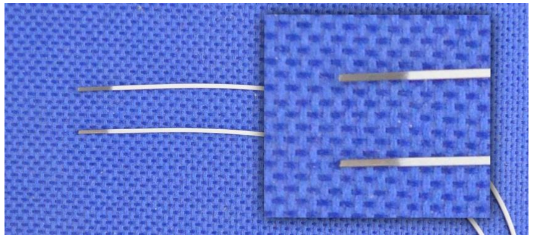

Figure 2.8 Uncoated part of wire

Then the remaining coated wire was cut from the most straight posterior part to a length of 30 mm with the use of a ruler by a wire cutter then they were marked with a permanent marker at 10,15 and 20 mm **(Fig 2.9).**

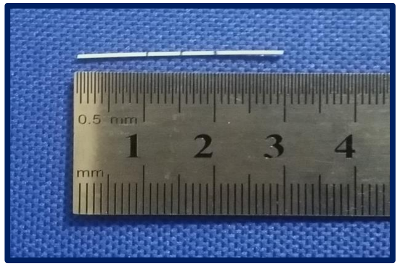

Figure 2.9 single marked specimen

A custom-made fulcrum was manufactured specially for this study by the researcher according to the ISO 15841 standard dimensions **(Fig 2.10)**.

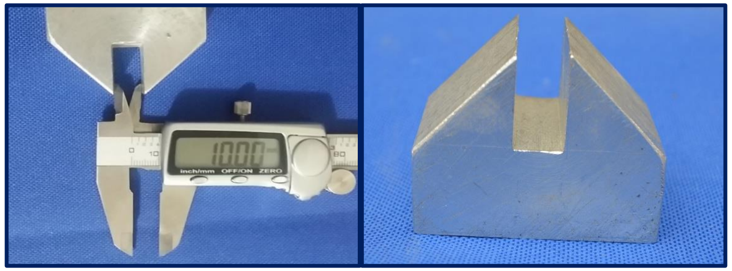

Figure 2.10: Custom manufactured fulcrum with the required dimensions

For frictional resistance test, 70 coated and 10 control wire samples involving both dimensions (0.016'' × 0.022'' and 0.019'' × 0.025'') were prepared from the pre-curved wires, with the use of ruler and a wire cutter. a 50 mm length of wire section was cut as a specimen **(Muguruma, 2017; Usui, 2017)**.

A group of 48 ceramic bracket (supplied from HUBIT company) with a 0.022 slot for the maxillary right premolar were selected for the test.

A total of 16 aluminum block (one for each brand size) with a dimensions of 40×15×9 mm where used which was custom-made **(Fig 2.11)** specially for the test by the author.

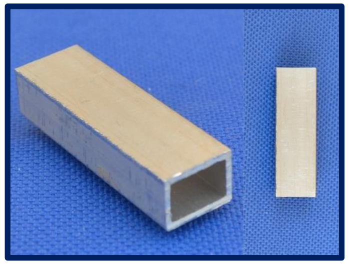

Figure 2.11: Aluminum blocks

With the use of bracket holder and cyanoacrylate adhesive (AMIR), every three brackets were mounted to single aluminum block in a straight line with 8 mm inter-bracket distance with the aid of a custom-made plastic template which was designed especially for this study **(Fig 2.12)**, and a straight wire segment of 0.0215''× 0.025'' so it was easier to exactly reproduce the same locations and angles of brackets **(Fig 2.13).**

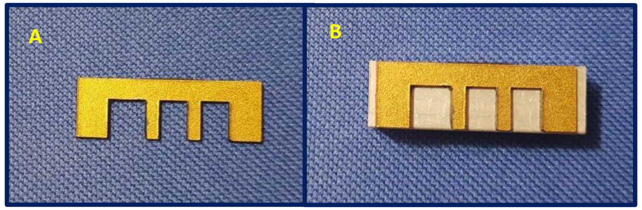

Figure 2.12 (A) Plastic template, (B) Template-aluminum block combination.

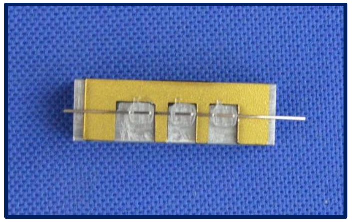

Figure 2.13 The setting of brackets

Every group of wires were contained in a certain package for adequate containment and protection from contamination during transit and storage in accordance with acceptable commercial practice and each package were labelled with the following information:

a) Name and address of the manufacturer

b) Name or trade name of wire

c) Dimensions of wire, in millimeters

d) The quantity of wires in number and length

2.2.3 Measurements of wire dimensions

Total wire dimension, inner core dimension, and thickness of coating layer of ninety-six wires (twelve for each brand, six for each dimension) were verified and measured by the metallurgical incident light microscope with an accuracy of ± 0.5 micrometer and 10 times magnification.

All specimens were measured from both wire aspects (width and height) by placing the wire segment under the microscope lens **(Fig 2.14 A)** and determine the dimensions using the attached computerized software which read the size in micrometer unit after determining the distance (red line) between both edges of wire aspect **(Fig 2.14 B)**.

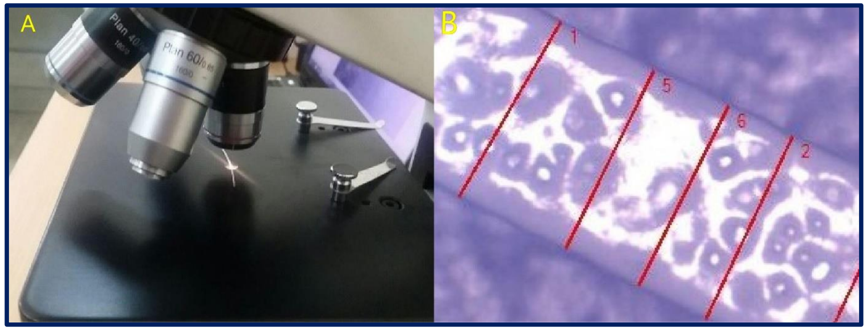

Figure 2.14 (A) Wire segment under microscope lens, (A) wire as it appeared in the computer display

Thickness of coating layer was measured after burning the wire segment by the torch for a duration of about 1-2 seconds then the coating was removed with a clean tissue paper, then measurements were done for the inner core stainless steel part and the thickness of the coating layer was measured by subtracting the inner wire dimension from the total dimension **(Fig 2.15)**.

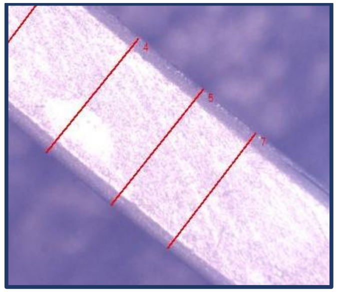

Figure 2.15 Burned wire under microscope

Measurements of wire dimension and thickness of coating layer is a new method and it was developed for the first time by the researcher because measurements in digital micrometer was not reliable.

2.2.4 Three-point-bending test

A computerized Instron H50KT Tinius Olsen testing machine (England) with a 10 N load cell was used for the experiments in the Ministry of Science and Technology where it was properly maintained and calibrated prior to testing. The machine consists of upper and lower jaws; the fulcrum was attached to the lower jaw while the intender was screwed to the upper movable part of the machine **(Fig 2.16)**.

Figure 2.16 Wire segment within the machine

The specification of the experiments was set according to the following parameters related to the ISO 15841 standard:

- The crosshead rate was 2 mm/min.
- The wires were subjected to a symmetrical three-point bending test.
- A span of wire 10 mm between supports was used and radi of fulcrum and intender was 0.1±0.05 mm **(Fig 2.17)**
- Deflection was carried out with a centrally-placed indenter.
- The test was performed in the direction of the height of the wire.
- The wires were deflected to a minimum permanent deflection of 2 mm
- Experiments were performed at room temperature 23±2°C.

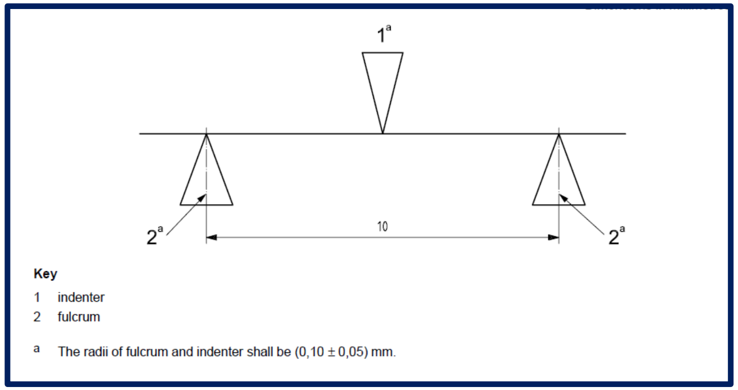

Figure 2.17: Three-point bending test (ISO standard, 2014)

Every wire segment was fixed onto the fulcrum with the help of the marked points **(Fig 2.18 A)**. Then by a computer-controlled stepper motor, loading was achieved through movement of a metal loading device (intender) adapted on the machine downward to the center of the wire and fulcrum to start bending test till a permanent deflection of a minimum of 2 mm was reached **(Fig 2.18 B)**.

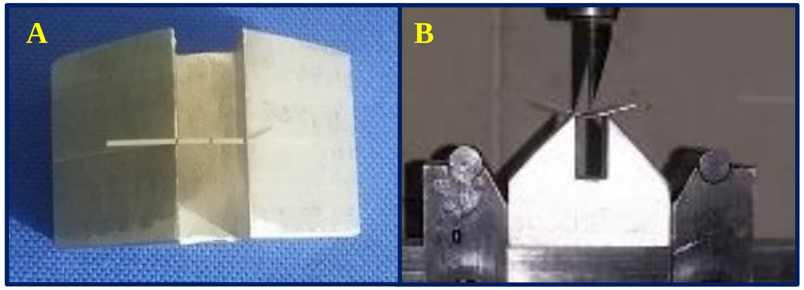

Figure 2.18: Loading of wire on fulcrum referenced by the marked points(A), bending test with wire-fulcrum combination (B).

A force- deflection **(Fig 2.19)** and stress- strain curve **(Fig 2.20)** were then plotted in the attached computer with the required following measurements:

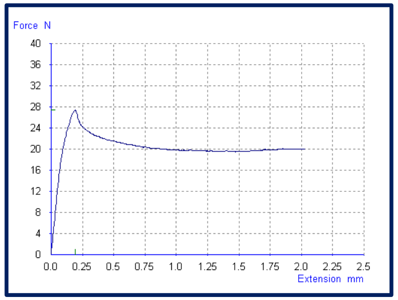

Figure 2.19 force deflection curve

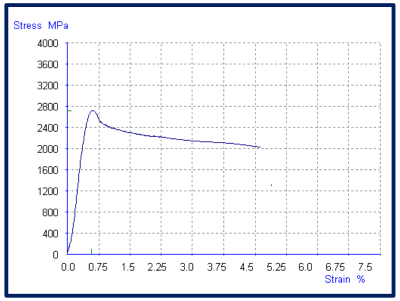

Figure 2.20 stress strain curve

- The yield strength (MPa) was measured directly from the stress strain curve in the computer at 0.2% strain offset.
- The modulus of elasticity (GPa) was calculated from the linear portion of force-deflection curves of each specimen using the equation **(Zweben *et al*.,1979; da *et al*. 2013):**

$$E = \frac{PL^3}{4BH^3D}$$

where P is load, L is span length, B is wire width, H is wire height, and D is deflection.

- Spring-back (SB) was calculated from the ratio of the yield strength (YS) to the modulus of elasticity (E):

$$SB = \frac{\text{YS}}{\text{E}}$$

2.2.5 Frictional resistance test

Eighty segments of wires (ten for each brand, five for each dimension) were ligated to the brackets using a ligature elastic with the use of an artery forceps **(Fig 2.21)**. The ligation was done with the use of hand gloves and tweezer to prevent contamination of the wires and brackets.

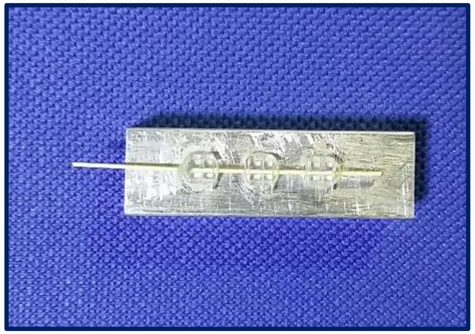

Figure 2.21: Wire-bracket-block system

A tensile test was used for this experiment by the universal testing machine (Instron). The aluminum block with the adhered bracket and ligated wire was gripped firmly by the lower head of the testing machine, while the end of the wire was attached directly to the clamp of the 10 N load cell located at the upper end of the testing machine **(Fig 2.22).**

Figure 2.22: Wire-bracket-block system fixed to machine

The specification of this test was used **(Hussain and Kumar, 2011; Kim *et al*., 2014; Muguruma, 2017)** as follow:

- The crosshead speed was 5 mm/min.
- The wire was pulled through a distance of 5 mm.
- For every group of wires two bracket-block combination were used (every block was used five times) to exclude any expected wearing of brackets.
- The wires were used only once, and the tests were carried out by the same investigator.
- All measurements were performed under dry conditions at room temperature.

The computer connected to the machine displayed a force-extension curve **(Fig 2.23)**, the maximum force value obtained as the wire moved across the distance was measured and recorded in Newton which was then converted into gram (gm) unit, and this represent the maximum frictional resistance force noted at the beginning of the movement (static friction).

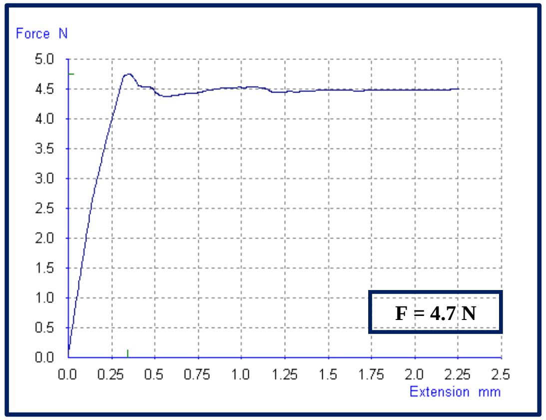

Fig 2.23: Force-extension curve with recorded maximum force

2.3 Pilot study

Initially, measurements of coating layer thickness were done by two methods which lead exactly to same results as the followings:

1- The coating layer was removed by scratching the coated surface of a wire segment only from one side by the edge of a metal rule. Then the thickness of coating from the other sides was viewed and measured **(Fig 2.24).** This is done from both wire aspects (width and height).

Figure 2.24 Thickness of coating layer

2- By burning the wire as it is explained previously, and this method was selected as the final procedure to exclude any probability of removing the inner stainless steel wire by scratching.

for three-point bending and frictional resistance tests, a pilot study with many trails was done prior to the final procedure to:

- test the workability and efficiency of the testing unit.
- Evaluate the feasibility of wire positioning of the three-point bending test
- Adjust proper software specification to reach the proper readings with training on the procedure, measurements and standardization.

2.4 Statistical Analysis

Data were collected, processed and analyzed by using statistical package of social science (SPSS) software version 20 with the following statistics:

1. Descriptive statistics:

- Shapiro-Wilks test for the distribution normality of data.
- Arithmetic mean.
- Standard deviation (SD).
- Minimum (Min.) and maximum (Max.).
- Figures and tables.

2. Inferential statistics:

- One-way analysis of variance (ANOVA) test.
- Least significant difference (LSD) test when ANOVA test was statistically significant.

Chapter Three

Results

Chapter Three

Results

The data obtained from the present experimental study were managed statistically to compare and explain the mechanical properties differences of different brans of aesthetic stainless steel arch wires.

3.1 Shapiro-Wilks test for normality of data distribution

It was found that all data of this experimental study including the five properties were normally distributed **(Table 3.1)** because the p- value of Shapiro-Wilk test is greater than 0.05 which mean non-significant**.**

Table 3.1: Shapiro-Wilks test for normality of distribution for the data of mechanical properties of both wire dimensions

Mechanical properties	Maximum Force		Yield strength		Elastic Modulus		Spring-back		Frictional Force	
Brand Name \ Dimension	0.016×0.022	0.019×0.022	0.016×0.022	0.019×0.022	0.016×0.022	0.019×0.022	0.016×0.022	0.019×0.022	0.016×0.022	0.019×0.022
IOS	.582	.647	.867	.804	.723	.076	.701	.257	.925	.158
DANY	.440	.963	.682	.942	.586	.577	.781	.940	.855	.084
DB	.716	.536	.181	.213	.609	.310	.411	.942	.774	.334
GH	.397	.360	.520	.352	.523	.430	.717	.667	.337	.813
Highland	.485	.059	.680	.266	.902	.573	.099	.264	.894	.096
Hubit	.895	.059	.875	.954	.855	.395	.380	.680	.695	.208
RMO	.422	.736	.378	.166	.772	.307	.269	.535	.918	.998
TP	.888	.220	.505	.739	.991	.224	.474	.296	.657	.201

d.f. = 6 for all properties data except for frictional force data d.f. = 5

3.2 Wire dimensions

Table 3.2 and 3.3 illustrate the mean and SD of the measured wire dimensions (width and height) with and without coating material (thickness of inner stainless steel wire). Table 3.4 show the thickness of coating layer for each type of coated wire.

For the fully coated wires, DANY showed a stainless steel inner core slightly larger than the stated sizes (0.1 to 0.2 mil). This was coated by 0.3 to 0.6 mil coating on all sides making the total wire larger than the stated sizes by 0.9 to 1.3 mil. While, Hubit wires, showed a stainless steel inner core slightly smaller than the stated sizes (0.2 to 0.9 mil). This was coated by 0.3 mil coating on all sides making the total wire comparable to the stated sizes from -0.3 to +0.4 mil.

On the other hand, GH and Highland wires showed a stainless steel inner core markedly smaller than the stated sizes (2 to 2.3 mil for GH and 2.1 to 2.3 mil for Highland) being closest for the width of the 0.019x0.025 wires. This was coated by 0.44 to 0.65 mil coating for GH and 0.35 to 0.5 mil coating for Highland wires on all sides making the total wire generally smaller than the stated sizes (especially for the height) by 0.7 to 1.6 mil for GH and 1.1 to 1.6 mil for Highland except for the width of the 0.019x0.025 wires.

For the labial coated wires, the coating only affected the width. Their height was very comparable to the stated sizes (-0.2 to +0.2 mil). For the width, the inner stainless steel core dimensions were also very close to the stated sizes (-0.2 to +0.3 mil). This was coated by a relatively thick labial coating (1.0 to 1.3 mil) making the width of the wire larger than the stated sizes by 1.0 to 1.4 mil.

Table 3.2: Mean and SD of measured total wire dimensions for 0.016''×0.022'' and 0.019''×0.025'' arch wires (measured in mil).

	0.016''×0.022''		0.019''×0.025''	
	Width	Height	Width	Height
Control	**22.0** ± 0.052	**15.8** ± 0.034	**24.9**± 0.012	**18.9**± 0.039
DANY	**23.3** ± 0.029	**16.9** ± 0.020	**26.2**± 0.038	**20.0**± 0.017
GH	**21.3** ± 0.023	**14.4** ± 0.035	**25.4**± 0.014	**17.8**± 0.038
Highland	**20.9** ± 0.044	**14.4** ± 0.020	**25.0**± 0.013	**17.6**± 0.007
Hubit	**21.8** ± 0.076	**15.7** ± 0.044	**25.1**± 0.019	**19.4**± 0.010
DB	**23.5** ± 0.078	**15.8**± 0.012	**26.1**± 0.008	**18.9**± 0.006
RMO	**22.8** ± 0.20	**16.0** ± 0.061	**26.1**± 0.021	**19.2**± 0.008
TP	**23.3** ± 0.110	**15.8** ± 0.074	**26.3** ± 0.007	**18.7**± 0.007

Measurements less and more than 1 mil of the stated dimensions are highlighted in red.

*** Wires has only labial coating**

Table 3.3: Mean and SD of measured inner wire dimensions for 0.016''×0.022'' and 0.019''×0.025'' arch wires (measured in mil).

	0.016''×0.022''		0.019''×0.025''	
	Width	Height	Width	Height
DANY	**22.0** ± 0.010	**16.2** ± 0.013	**25.0** ± 0.010	**19.1** ± 0.024
GH	**20.0** ± 0.066	**13.7** ± 0.012	**24.5** ± 0.017	**16.8** ± 0.012
Highland	**19.9** ± 0.032	**13.7** ± 0.019	**24.5** ± 0.026	**16.7** ± 0.010
Hubit	**21.3** ± 0.014	**15.1** ± 0.022	**24.6** ± 0.018	**18.8** ± 0.02
DB*	**22.3** ± 0.040	-	**25.0**± 0.020	-
RMO*	**21.8**± 0.003	-	**24.9** ± 0.003	-
TP*	**21.9**± 0.012	-	**24.9** ± 0.009	-

Measurements less than 2 mil of the stated dimensions are highlighted in red.

*** Wires has only labial coating**

Table 3.4 Mean and S.D for the thickness of coating layers for 0.016'' × 0.022'' and 0.019''× 0.025'' in width (labial and lingual) and height (occlusal and gingival).

	0.016''×0.022''				0.019''×0.025''			
	Width		Height		Width		Height	
	mil	µm	mil	µm	mil	µm	mil	µm
DANY	0.636 ±0.016	16.172 ±0.407	0.341 ±0.007	8.685 ±0.170	0.567 ±0.019	14.440 ±0.471	0.391 ±0.014	9.944 ±0.349
GH	0.658 ±0.092	16.746 ±2.333	0.375 ±0.023	9.551 ±0.581	0.446 ±0.012	11.337 ±0.317	0.490 ±0.019	12.477 ±0.489
Highland	0.506 ±0.035	12.876 ±0.888	0.353 ±0.004	8.972 ±0.107	0.468 ±0.017	11.917 ±0.431	0.426 ±0.011	10.841 ±0.290
Hubit	0.293 ±0.043	7.468 ±1.096	0.301 ±0.019	7.468 ±0.491	0.256 0.014	6.522 ±0.354	0.317 ±0.009	8.077 ±0.225
DB*	1.248 ±0.051	31.759 ±1.294	-	-	1.056 ±0.029	26.861 ±0.746	-	-
RMO*	0.943 ±0.077	23.988 ±1.969	-	-	1.161 ±0.022	29.536 ±0.551	-	-
TP*	1.412 ±0.111	35.924 ±2.824	-	-	1.350 ±0.014	34.345 ±0.368	-	-

*** Only has labial coating**

3.3 Mechanical properties from three-point bending test

3.3.1 Maximum deflection force

Figure 3.1 and 3.2 revealed the mean values of maximum deflection force for both 0.016''×0.022'' and 0.019''×0.025 arch wires. ANOVA test for both wire dimensions showed a highly significant difference between wires (Table 3.5 and 3.6). LSD test was performed for comparison between each two types of arch wires and the results are displayed in table 3.7.

The maximum deflection force was highest for the labially coated DB wires followed by the fully coated DANY wires being significantly larger than the uncoated control wires for both wire dimensions with no significant difference between DB and DANY wires (Table 3.7).

This was followed by the two labially coated wires (RMO and TP) which were very comparable to each other and to the uncoated control wires for both wire dimensions being statistically insignificant among them (Table 3.7).

The lowest maximum deflection force was noted in the fully coated Hubit, Highland and GH wires in descending order. These readings were significantly smaller than that of the uncoated control wires for both wire dimensions and showed statistically significant differences between the three wires (Table 3.7).

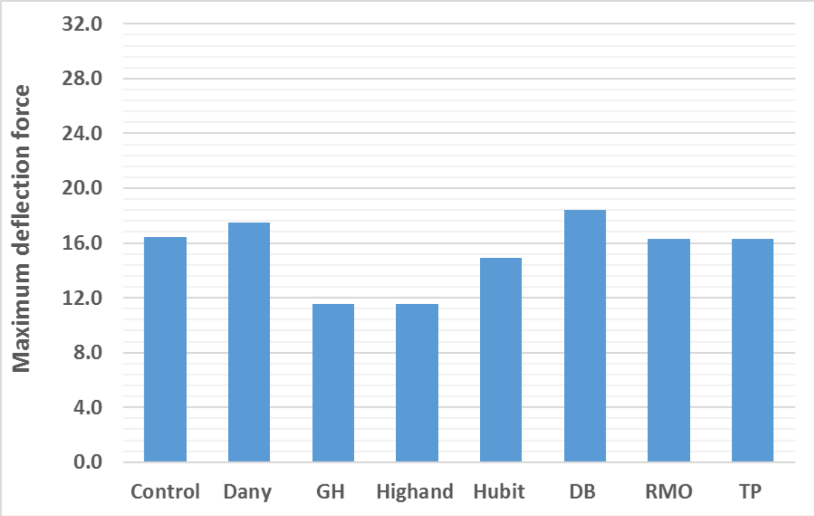

Figure 3.1: Mean maximum deflection force (N) of 0.016''×0.022'' wires.

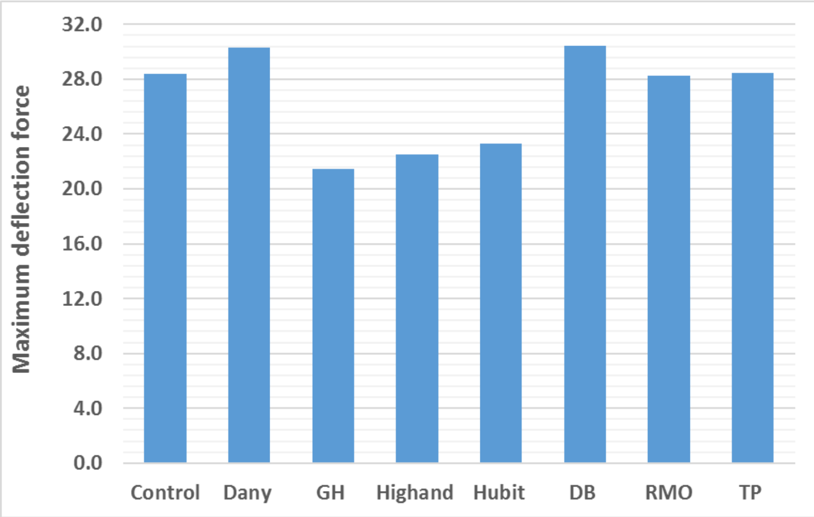

Figure 3.2: Mean maximum deflection force (N) of 0.019''×0.025'' wires.

Table 3.5: Descriptive statistics and ANOVA tests for the maximum deflection force (N) for 0.016''×0.022'' arch wires

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	6	16.410	0.278	16.09	16.78	827.723	0.000
DANY	6	17.483	0.210	17.13	17.7		
GH	6	11.518	0.173	11.29	11.72		
Highland	6	11.562	0.273	11.08	11.87		
Hubit	6	14.910	0.175	14.65	15.13		
DB	6	17.857	0.286	17.41	18.16		
RMO	6	16.283	0.226	16	16.54		
TP	6	16.327	0.199	16.08	16.62		

Table 3.6: Descriptive statistics and ANOVA tests for the maximum deflection force (N) for 0.019''×0.025'' arch wires

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	6	28.370	0.249	28.09	28.74	1126.737	0.000
DANY	6	30.275	0.204	29.98	30.54		
GH	6	21.461	0.228	21.07	21.67		
Highland	6	22.503	0.357	22	22.82		
Hubit	6	23.326	0.315	23	23.65		
DB	6	30.413	0.282	30.09	30.78		
RMO	6	28.281	0.211	28.04	28.58		
TP	6	28.46	0.219	28.07	28.65		

Table 3.7 Differences of maximum deflection force (LSD test).

		0.016''×0.022''		0.019''×0.025''	
		Mean difference	p-value	Mean difference	p-value
Control	DANY	-1.073	0.000	-1.905	0.000
	GH	4.892	0.000	6.908	0.000
	Highland	4.848	0.000	5.867	0.000
	Hubit	1.500	0.000	5.043	0.000
	DB	-1.447	0.000	-2.043	0.000
	RMO	0.127	**0.978**	0.088	**0.999**
	TP	0.083	**0.998**	-0.090	**0.999**
DANY	GH	5.965	0.000	8.813	0.000
	Highland	5.922	0.000	7.772	0.000
	Hubit	2.573	0.000	6.948	0.000
	DB	-0.373	**0.120**	-0.138	**0.984**
	RMO	1.200	0.000	1.993	0.000
	TP	1.157	0.000	1.815	0.000
DB	GH	6.338	0.000	8.952	0.000
	Highland	6.295	0.000	7.910	0.000
	Hubit	2.947	0.000	7.087	0.000
	RMO	1.573	0.000	2.132	0.000
	TP	1.530	0.000	1.953	0.000
GH	Highland	-0.043	**1.000**	-1.042	0.000
	Hubit	-3.392	0.000	-1.865	0.000
	RMO	-4.765	0.000	-6.820	0.000
	TP	-4.808	0.000	-6.998	0.000
Highland	Hubit	-3.348	0.000	-0.823	0.000
	RMO	-4.722	0.000	-5.778	0.000
	TP	-4.765	0.000	-5.957	0.000
Hubit	RMO	-1.373	0.000	-4.955	0.000
	TP	-1.417	0.000	-5.133	0.000
RMO	TP	-0.043	**1.000**	-0.178	**0.935**

3.3.2 Yield strength

Figure 3.3 and 3.4 revealed the mean values of yield strength for both 0.016''×0.022'' and 0.019''×0.025 arch wires. The means for all wires of 0.016''×0.022'' arch wires ranged from 1382.3 MPa for the GH wires to 1880.33 MPa for DANY wires and from 1532.4 MPa (Highland wires) to 2204 MPa (DANY wires) for 0.019''×0.025'' arch wire.

ANOVA test for both wire dimensions showed a highly significant difference between wires (Table 3.8 and 3.9). LSD test was performed for comparison between each two types of arch wires and the results are displayed in table 3.10.

The yield strength was highest for the fully coated DANY wires followed by the labially coated DB wires being significantly larger than the uncoated control wires for both wire dimensions with no significant difference between DB and DANY wires in 0.019''×0.025'' wires (Table 3.10).

This was followed by the two labially coated wires (RMO and TP) and the fully coated Hubit wires which were very comparable to each other and to the uncoated control wires for both wire dimensions being statistically insignificant among them (Table 3.10).

The lowest yield strength was noted in the fully coated Highland and GH wires. These readings were significantly smaller than that of the uncoated control wires for both wire dimensions and showed statistically insignificant differences between them (Table 3.10).

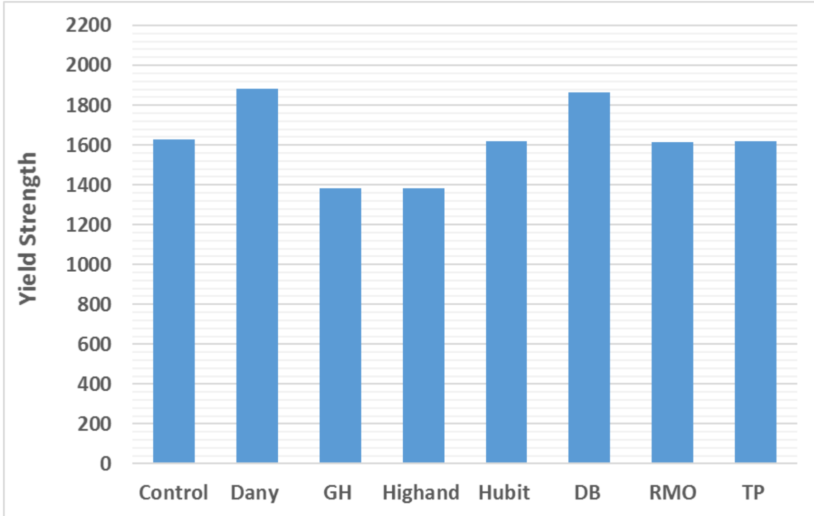

Figure 3.3: Means for the yield strength (MPa) of 0.016''×0.022'' wires.

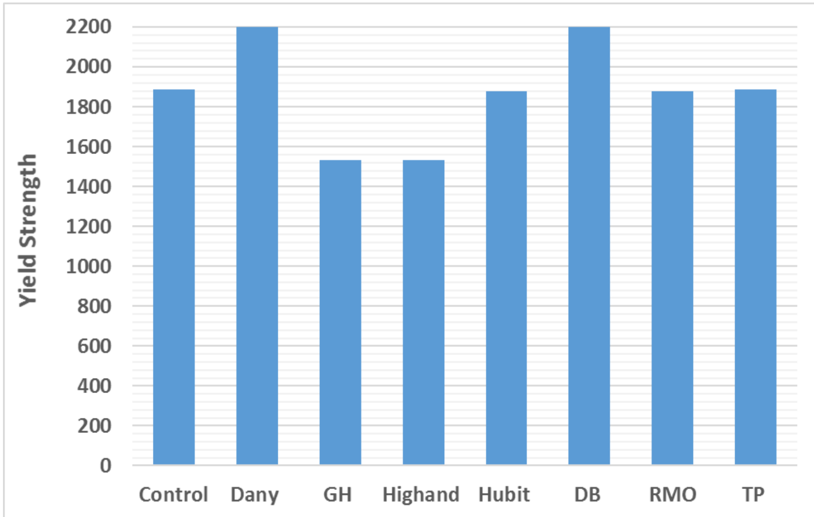

Figure 3.4: Means for the yield strength (MPa) of 0.019''×0.025'' wires

Table 3.8: Descriptive statistics and ANOVA tests for the yield strength of 0.016''×0.022'' arch wires

	Descriptive statistics					Comparison	
	N	**Mean (MPa)**	**SD**	**Min.**	**Max.**	**F-test**	**p-value**
Control	**6**	1627.500	3.563	1622	1632	**4870.891**	**0.000**
DANY	**6**	1880.333	5.853	1872	1888		
GH	**6**	1382.333	4.033	1378	1388		
Highland	**6**	1384.333	5.006	1379	1392		
Hubit	**6**	1619.667	4.273	1614	1626		
DB	**6**	1862.000	3.286	1859	1868		
RMO	**6**	1616.333	5.125	1610	1622		
TP	**6**	1618.667	3.265	1614	1622		

Table 3.9: Descriptive statistics and ANOVA tests for the yield strength for 0.019''×0.025'' arch wires

	Descriptive statistics					Comparison	
	N	**Mean (MPa)**	**SD**	**Min.**	**Max.**	**F-test**	**p-value**
Control	**6**	1887.200	2.280	1884	1890	**10010.240**	**0.000**
DANY	**6**	2204.000	4.949	2197	2210		
GH	**6**	1533.600	7.536	1522	1540		
Highland	**6**	1532.400	11.52	1520	1544		
Hubit	**6**	1876.000	4.732	1870	1882		
DB	**6**	2200.800	6.099	2194	2208		
RMO	**6**	1876.200	4.381	1871	1880		
TP	**6**	1886.800	3.033	1882	1890		

Table 3.10: Differences in yield strength for each wire type using LSD test

		0.016''×0.022''		0.019''×0.025''	
		Mean difference	**p-value**	**Mean difference**	**p-value**
Control	**DANY**	-252.833	0.000	-316.800	0.000
	GH	245.167	0.000	353.600	0.000
	Highland	243.167	0.000	354.800	0.000
	Hubit	7.833	**0.435**	11.200	**0.059**
	DB	-234.500	0.000	-313.600	0.000
	RMO	11.167	**0.435**	11.000	**0.067**
	TP	8.833	**0.287**	0.400	**1.000**
DANY	**GH**	498.000	0.000	670.400	0.000
	Highland	496.000	0.000	671.600	0.000
	Hubit	260.667	0.000	328.000	0.000
	DB	18.333	0.000	3.200	**0.985**
	RMO	264.000	0.000	327.800	0.000
	TP	261.667	0.000	317.200	0.000
DB	**GH**	479.667	0.000	667.200	0.000
	Highland	477.667	0.000	668.400	0.000
	Hubit	242.333	0.000	324.800	0.000
	RMO	245.667	0.000	324.600	0.000
	TP	243.333	0.000	314.000	0.000
GH	**Highland**	-2.000	**0.999**	1.200	**1.000**
	Hubit	-237.333	0.000	-342.400	0.000
	RMO	-234.000	0.000	-342.600	0.000
	TP	-236.333	0.000	-353.200	0.000
Highland	**Hubit**	-235.333	0.000	-343.600	0.000
	RMO	-232.000	0.000	-343.800	0.000
	TP	-234.333	0.000	-354.400	0.000
Hubit	**RMO**	3.333	**1.000**	-0.200	**1.000**
	TP	1.000	**1.000**	-10.800	**0.076**
RMO	**TP**	-2.333	**1.000**	-10.600	**0.087**

3.3.3 Modulus of elasticity

Figure 3.5 and 3.6 revealed the mean values of modulus of elasticity for both 0.016''×0.022'' and 0.019''×0.025 arch wires. The means for all wires of 0.016''×0.022'' arch wires ranged from 120.4 GPa for the GH wires to 180.3 GPa (DB) and from 141.6 GPa (GH) to 192.9 GPa (DB) for 0.019''×0.025'' arch wire.

ANOVA test for both wire dimensions showed a highly significant difference between wires (Table 3.11 and 3.12). LSD test was performed for comparison between each two types of arch wires and the results are displayed in table 3.13.

The modulus of elasticity was highest for the labially coated DB wires followed by the fully coated DANY wires being significantly larger than the uncoated control wires for both wire dimensions with a significant difference between DB and DANY wires (Table 3.13).

This was followed by the two labially coated wires (RMO and TP) which were very comparable to each other and to the uncoated control wires for both wire dimensions being statistically insignificant among them (Table 3.13).

The lowest modulus of elasticity was noted in the fully coated Hubit, Highland and GH wires in descending order. These readings were significantly smaller than that of the uncoated control wires for both wire dimensions and showed statistically significant differences between the three wires with Hubit being noticeably higher than Highland and GH wires (Table 3.13).

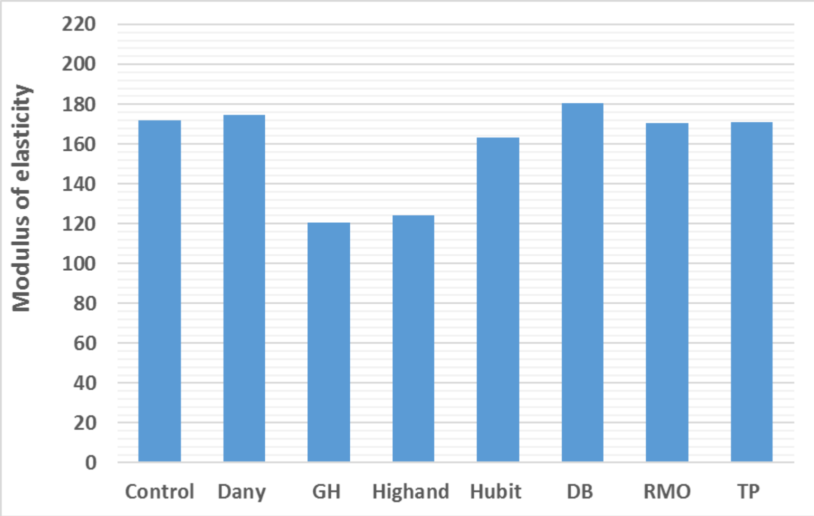

Figure 3.5: Mean modulus of elasticity (GPa) of 0.016''×0.022'' wires.

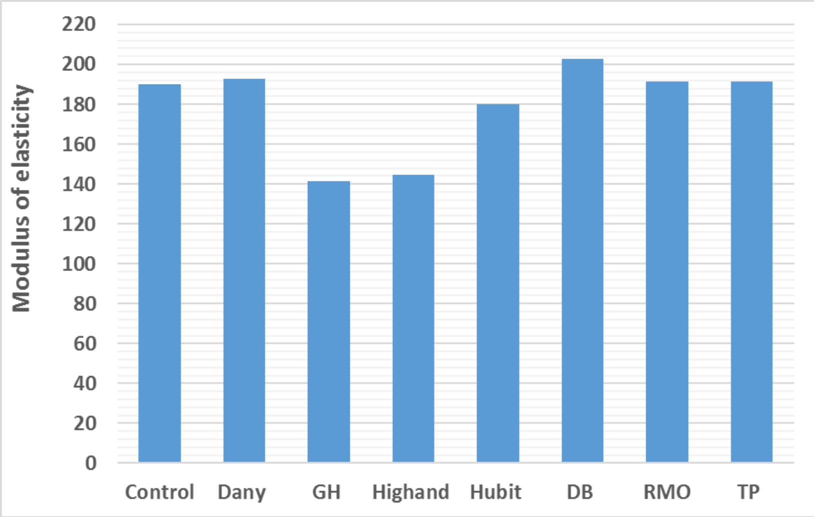

Figure 3.6: Means modulus of elasticity (GPa) of 0.019''×0.025'' wires.

Table 3.11: Descriptive statistics and ANOVA tests for the modulus of elasticity for 0.016''×0.022'' arch wires

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	**6**	171.993	1.033	170.77	173.56	**2316.802**	**0.000**
DANY	**6**	174.631	0.973	173.55	176.33		
GH	**6**	120.435	1.454	118.65	122.14		
Highland	**6**	124.271	1.122	122.88	125.87		
Hubit	**6**	163.355	1.452	161.57	165.39		
DB	**6**	180.328	1.004	179.00	181.56		
RMO	**6**	170.725	1.498	168.49	172.51		
TP	**6**	170.848	0.811	169.77	172.00		

Table 3.12: Descriptive statistics and ANOVA tests for the modulus of elasticity for 0.019''×0.025'' arch wires.

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	**6**	189.986	1.129	189.00	192.14	**2768.992**	**0.000**
DANY	**6**	192.931	0.882	191.77	194.00		
GH	**6**	141.555	1.368	140.00	144.00		
Highland	**6**	144.405	1.455	142.88	146.78		
Hubit	**6**	179.978	0.407	179.53	180.53		
DB	**6**	202.821	1.486	201.00	204.66		
RMO	**6**	191.460	0.830	190.55	192.55		
TP	**6**	191.505	0.530	190.77	192.04		

Table 3.13: Differences in modulus of elasticity using LSD test.

		0.016''×0.022''		0.019''×0.025''	
		Mean difference	p-value	Mean difference	p-value
Control	**DANY**	-2.638	0.000	-2.945	0.000
	GH	51.558	0.000	48.432	0.000
	Highland	47.722	0.000	45.582	0.000
	Hubit	8.638	0.000	10.008	0.000
	DB	-8.335	0.000	-12.835	0.000
	RMO	1.268	**0.598**	-1.473	**0.291**
	TP	1.145	**0.712**	-1.518	**0.257**
DANY	**GH**	54.197	0.000	51.377	0.000
	Highland	50.360	0.000	48.527	0.000
	Hubit	11.277	0.000	12.953	0.000
	DB	-5.697	0.000	-9.890	0.000
	RMO	3.907	0.000	1.472	**0.292**
	TP	3.783	0.000	1.427	**0.329**
DB	**GH**	59.893	0.000	61.267	0.000
	Highland	56.057	0.000	58.417	0.000
	Hubit	16.973	0.000	22.843	0.000
	RMO	9.603	0.000	11.362	0.000
	TP	9.480	0.000	11.317	0.000
GH	**Highland**	-3.837	0.000	-2.850	0.000
	Hubit	-42.920	0.000	-38.423	0.000
	RMO	-50.290	0.000	-49.905	0.000
	TP	-50.413	0.000	-49.950	0.000
Highland	**Hubit**	-39.083	0.000	-35.573	0.000
	RMO	-46.453	0.000	-49.905	0.000
	TP	-46.577	0.000	-49.950	0.000
Hubit	**RMO**	-7.370	0.000	0.000	0.000
	TP	-7.493	0.000	-11.482	0.000
RMO	**TP**	-0.123	**1.000**	-0.045	**1.000**

3.3.4 Spring-back

Figure 3.7 and 3.8 revealed the mean spring-back values for both 0.016''×0.022'' and 0.019''×0.025 arch wires. The means for all wires of 0.016''×0.022'' wires ranged from 0.0094 for the TP, RMO and control wires to 0.0114 (GH wires) and from 0.0098 (TP and RMO wires) to 0.0114 (DANY wires) for 0.019''×0.025'' wires.

ANOVA test for both wire dimensions showed a highly significant difference between wires (Table 3.14 and 3.15). LSD test was performed for comparison between each two types of arch wires and the results are displayed in table 3.16.

The spring-back of GH, Highland, Dany, DB and Hubit wires were all significantly higher than the uncoated control wires for both wire dimensions with significant difference between the five wire types (Table 3.16).

However, the lowest spring-back values were noted in the labially coated TP and RMO wires. These readings very comparable to the uncoated control wires for both wire dimensions and showed a statistically insignificant differences between them (Table 3.16).

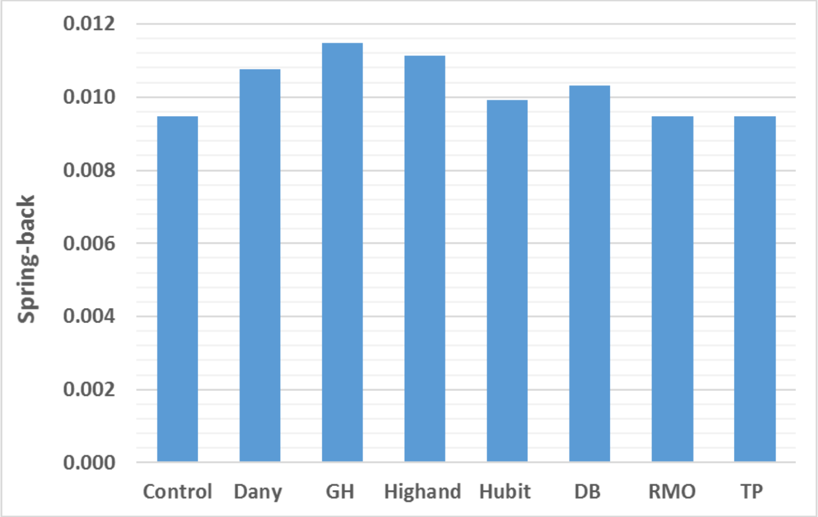

Figure 3.7: Means of the spring-back of 0.016''×0.022'' wires.

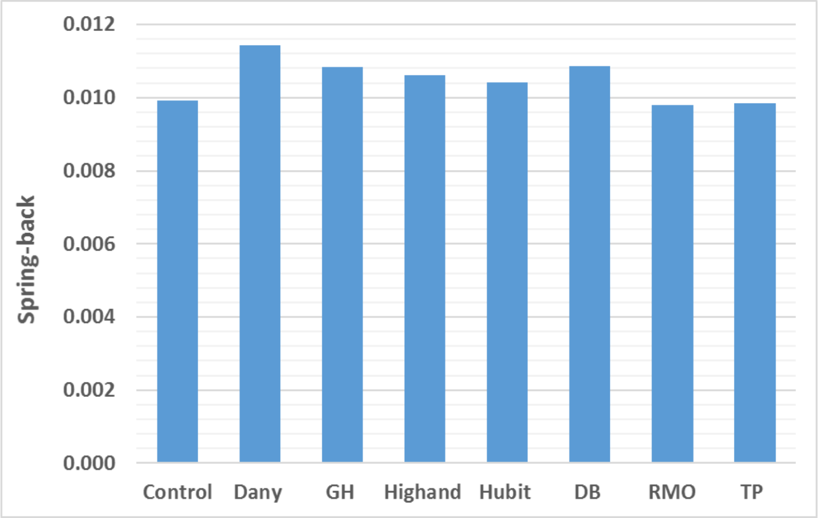

Figure 3.8: Means of the spring-back of 0.019''×0.025'' wires.

Table 3.14: Descriptive statistics and ANOVA tests for the spring-back for 0.016''×0.022'' arch wires

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	6	0.0094	5.3×10^{-5}	0.0094	0.0095	557.464	0.000
DANY	6	0.0107	7×10^{-5}	0.0106	0.0108		
GH	6	0.0114	14×10^{-5}	0.0112	0.0116		
Highland	6	0.0111	9×10^{-5}	0.0110	0.0112		
Hubit	6	0.0099	6×10^{-5}	0.0098	0.0099		
DB	6	0.0103	4×10^{-5}	0.0102	0.0104		
RMO	6	0.0094	8×10^{-5}	0.0093	0.0096		
TP	6	0.0094	5×10^{-5}	0.0093	0.0095		

Table 3.15: Descriptive statistics and ANOVA tests for the spring-back for 0.019''×0.025'' arch wires

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	6	0.0099	6×10^{-5}	0.00981	0.00998	293.586	0.000
DANY	6	0.0114	5×10^{-5}	0.01133	0.01149		
GH	6	0.0108	9×10^{-5}	0.01069	0.01098		
Highland	6	0.0106	15×10^{-5}	0.01036	0.01076		
Hubit	6	0.0104	4×10^{-5}	0.01035	0.01048		
DB	6	0.0108	9×10^{-5}	0.01073	0.01098		
RMO	6	0.0098	4×10^{-5}	0.00971	0.00985		
TP	6	0.0098	3.2×10^{-5}	0.00979	0.00987		

Table 3.16: Differences in spring-back using LSD test

		0.016''×0.022''		0.019''×0.025''	
		Mean difference	p-value	Mean difference	p-value
Control	DANY	-0.001	0.000	-0.001	0.000
	GH	-0.002	0.000	-0.001	0.000
	Highland	-0.002	0.000	-0.001	0.000
	Hubit	0.000	0.000	0.000	0.000
	DB	-0.001	0.000	-0.001	0.000
	RMO	0.000	**1.000**	0.000	**0.148**
	TP	0.000	**1.000**	0.000	**0.664**
DANY	GH	-0.001	0.000	0.001	0.000
	Highland	0.000	0.000	0.001	0.000
	Hubit	0.001	0.000	0.001	0.000
	DB	0.000	0.000	0.001	0.000
	RMO	0.001	**0.305**	0.002	0.000
	TP	0.001	**0.288**	0.002	0.000
DB	GH	-0.001	0.000	0.000	**0.998**
	Highland	-0.001	0.000	0.000	0.000
	Hubit	0.000	0.000	0.000	0.000
	RMO	0.001	0.000	0.001	0.000
	TP	0.001	0.000	0.001	0.000
GH	Highland	0.000	0.000	0.000	0.002
	Hubit	0.002	0.000	0.000	0.000
	RMO	0.002	**0.995**	0.001	0.000
	TP	0.002	**0.995**	0.001	0.000
Highland	Hubit	0.001	0.000	0.000	0.004
	RMO	0.002	**0.945**	0.001	0.000
	TP	0.002	**0.945**	0.001	0.000
Hubit	RMO	0.000	0.000	0.001	0.000
	TP	0.000	0.000	0.001	0.000
RMO	TP	0.000	**1.000**	0.000	**0.975**

3.4 Frictional resistance force

Figure 3.9 and 3.10 revealed the mean values of static friction for both 0.016''×0.022'' and 0.019''×0.025 arch wires. The means for all wires of 0.016''×0.022'' arch wires ranged from 246.8 gm for Hubit wires to 464.0 gm for TP wires and from 344.0 gm (Hubit wires) to 534.8 MPa (TP wires) for 0.019''×0.025'' arch wire.

ANOVA test for both wire dimensions showed a highly significant difference between wires (Table 3.17 and 3.18). LSD test was performed for comparison between each two types of arch wires and the results are displayed in table 3.19.

Friction was higher for the labially coated wires TP, DB and RMO in descending order; being significantly larger than the uncoated control wires for both wire dimensions (Table 3.19).

However, the four fully coated wires (Dany, Highland, GH and Hubit in descending order) showed significantly less friction than the uncoated control wires for both wire dimensions being statistically significant among them except for GH and Highland which displayed a non-significant difference (Table 3.19).

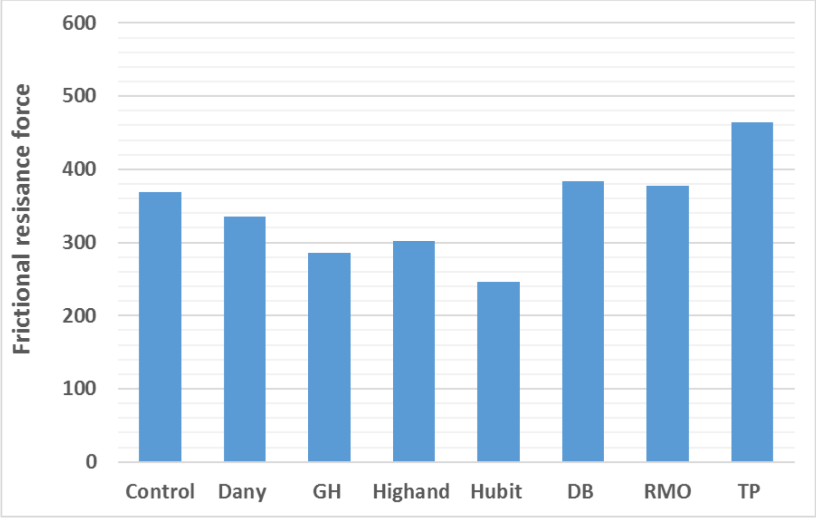

Figure 3.9: Mean changes for the frictional resistance force of 0.016''×0.022'' wires.

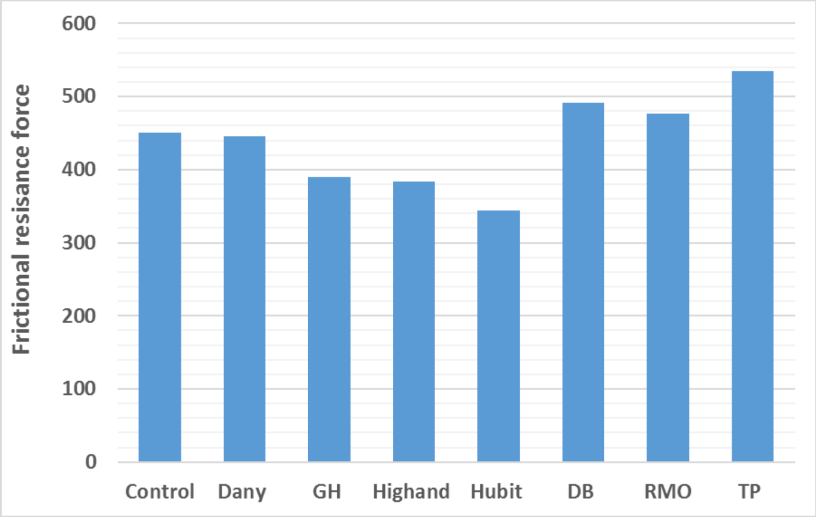

Figure 3.10: Mean changes for the frictional resistance force of 0.019''×0.025'' wires.

Table 3.17: Descriptive statistics and ANOVA tests for the static friction of 0.016''×0.022'' arch wires

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	5	368.621	9.668	355.875	381.367	205.4	0.000
DANY	5	335.481	14.025	313.047	354.855		
GH	5	286.365	7.856	276.3387	295.713		
Highland	5	301.831	12.236	284.496	318.146		
Hubit	5	246.767	8.889	234.531	257.984		
DB	5	384.256	9.596	368.111	395.643		
RMO	5	377.628	10.087	364.032	388.50		
TP	5	463.963	16.315	447.648	489.45		

Table 3.18: Descriptive statistics and ANOVA tests for the static friction for 0.019''×0.025'' arch wires

	Descriptive statistics					Comparison	
	N	Mean	SD	Min.	Max.	F-test	p-value
Control	5	450.197	5.971	441.530	457.845	176.2	0.000
DANY	5	446.118	15.951	426.234	467.022		
GH	5	389.695	14.838	368.111	407.88		
Highland	5	383.917	9.046	374.229	398.702		
Hubit	5	343.978	9.271	331.402	354.855		
DB	5	492.005	7.145	481.298	500.672		
RMO	5	476.199	5.945	468.040	484.350		
TP	5	534.832	18.066	502.710	550.630		

Table 3.19: Differences in frictional forces using LSD test.

		0.016''×0.022''		0.019''×0.025''	
		Mean difference	p-value	Mean difference	p-value
IOS	**DANY**	33.140	0.0003	4.078	**0.999**
	GH	82.255	0.000	60.502	0.000
	Highland	66.790	0.000	66.280	0.000
	Hubit	121.854	0.000	106.218	0.000
	DB	-15.635	**0.281**	-41.807	0.000
	RMO	-9.007	0.002	-26.002	0.009
	TP	-95.341	0.000	-84.635	0.000
DANY	**GH**	49.115	0.000	56.423	0.000
	Highland	33.650	0.000	62.201	0.000
	Hubit	88.713	0.000	102.139	0.000
	DB	-48.775	0.000	-45.886	0.000
	RMO	-42.147	0.000	-30.081	0.002
	TP	-128.482	0.000	-88.713	0.000
DB	**GH**	97.891	0.000	102.309	0.000
	Highland	82.4257	0.000	108.088	0.000
	Hubit	137.489	0.001	148.026	0.006
	RMO	6.628	0.000	15.805	**0.296**
	TP	-79.706	0.000	-42.827	0.000
GH	**Highland**	-15.465	**0.294**	5.778	**0.988**
	Hubit	39.598	0.000	45.716	0.000
	RMO	-91.263	0.000	-86.504	0.000
	TP	-177.597	0.000	-145.137	0.001
Highland	**Hubit**	55.063	0.000	39.938	0.000
	RMO	-75.797	0.000	-92.282	0.000
	TP	-162.132	0.000	-150.915	0.010
Hubit	**RMO**	-130.861	0.000	-132.221	0.000
	TP	-217.196	0.000	-190.853	0.000
RMO	**TP**	-86.334	0.000	-58.632	0.000

Chapter Four

Discussion

Chapter Four

Discussion

4.1 Study design

4.1.1 Sample selection

A total of 96 of wire segments (0.016'' × 0.022'' and 0.019'' × 0.025'') for the three-point bending test and 80 segment of wires for the frictional test were used **(ISO 15841, 2014; Muguruma, 2017)**. These wire dimension were selected because they are very common dimensions used in clinical orthodontic treatment and orthodontic literature.

4.1.2 Three-point bending test

In this study, a three-point bending test used a simple free ends beam theory and the deflection was measured at the mid portion of the wire segment according to the specification of the ISO 15841 standard.

The three-point bending test is identified as the most appropriate test for force-deflection analysis and allows one to analyze the bending forces for a given deflection for all of the various wires. It was performed due to its ability to closely simulate the clinical applications of arch wires and to assess and differentiate their force deflection properties **(Krishnan and Kumar, 2004).**

4.1.3 Frictional resistance test

There is a controversy concerning which frictional force is more significant in clinical orthodontic treatment. In the study by **Drescher *et al*. (1989)**, the static friction and kinetic friction occurred at nearly the same time owing to the low speed of orthodontic tooth movement, so distinguishing between the static and kinetic frictional forces was difficult.

Frank and Nikolai (1980) used the maximum kinetic frictional force, **Kusy and Whitley (1990)** measured both static and kinetic frictional forces, while **Tselepis *et al*. (1994)** used only the kinetic fictional force after the maximum static frictional force was reached.

In a study by **Burrow (2009)**, it was found that the static frictional force was more appropriate than the kinetic frictional force as orthodontic tooth movement is not continuous. For this reason, only the maximum static frictional force was measured in the present study.

In this study, a customized template was used which is made from hard plastic material and a segment of stainless steel wire of 0.0215''× 0.025'' to determine the exact locations and angles of brackets. The inter-bracket distance is equal to eight millimeters which is equivalent to the inter-bracket distance between maxillary premolars.

The cross head speed rate was chosen at 5 mm / minute in accordance with previous studies **(Hareeja, 2014; Kim *et al*., 2014; Aloysius, 2015; Gómez, 2016)**, and since different studies show no significant differences by changing speed from 0.5mm/minute to 50mm/minute **(Ireland *et al*., 1991; Taylor and Ison, 1996; Ryuji, 2007).**

It is difficult to standardize the magnitude of ligation force exerted when using stainless steel ligatures **(Plaza *et al*., 2010; Tada, 2017)**. To reduce the probability for such bias, elastic ligatures were used in this study.

4.1.4 Cross section dimensions of arch wires

None of the tested wires matched the stated dimensions given by the manufacturer with a trend showing smaller dimensions for fully coated wires (GH, Highland and Hubit) and larger width for labially coated wires (DB, RMO and TP) and the fully coated wire DANY. This inconsistency in wire dimensions

and inability of manufacturer to produce wires precisely have been previously reported by **Melling (1997) and da Silva *et al*. (2013).**

The fully coated arch wires (GH, Highland and Hubit) had smaller stainless steel inner cores to give space for the coating layer as most manufacturers state the wire size with the coating **(da Silva *et al*., 2013; Mugurumaa *et al*. 2017).** This was more evident in GH and Highland wires. This highlights the need for practitioners to make sure of the real dimension of the wires they clinically use.

The coating layer thickness differed among brands being of 0.3 to 0.6 mil in fully coated wires and 1.0 to 1.4 mil in labially coated wires. However, since fully coated wires are covered from the 4 sides, the relatively thinner coating takes more volume than the relatively thicker labial coating only, and this affects overall wire dimensions.

4.2 Elastic behavior of the wires

The maximum deflection forces, yield strength and the modulus of elasticity measured for the control uncoated and RMO and TP labially coated wires were almost the same with no significant differences between them for both wire dimensions since they kept the inner alloy core dimensions without compromising their properties.

However, the labial coated DB wires and fully coated DANY wires had nearly the exact inner stainless steel dimensions, but showed higher values of maximum deflection force, yield strength and elastic modulus. This may be attributed to a stiffer stainless steel inner wire core being either because of the chemical composition or work hardening. The manufacturers did not respond to our request to disclose this information as they regarded it as an industrial secret.

The fully epoxy coated arch wires from GH and Highland show lower deflection forces, lower yield strength with lower modulus of elasticity and this is strongly attributed to the decreased thickness of the inner stainless steel wire used to compensate for the thickness of the coating layer which result in lower mechanical properties. These arch wires will deliver low forces when activated in the oral cavity leading to inadequate control of tooth movements **(Peter *et al.*, 2012)**. Furthermore, the reduced values of yield strength of GH and Highland can lead to time-dependence force loss and deformation when deflected to an extent reaching the yield point **(Goldberg, 2010).** These arch wires are not expected to have clinical behavior similar to regular arch wires with the same dimensions and under same conditions.

On the other hand, Hubit wires show deflection force and elastic modulus slightly more than GH and Highland, but they show yield strength as high as the non-coated wires. This may be attributed to its greater inner alloy core dimensions together with different alloy composition.

These results agree with those reported by **da Silva *et al.* (2013) and Mugurumaa *et al*. (2017)** who concluded that the reduction of the inner alloy core dimensions for fully coated stainless steel wires greatly affects the mechanical properties of coated stainless steel wires with lower elastic modulus. While, the properties of labially coated wires according to **da Silva *et al.* (2013)** practically did not differ from their corresponding control wires with no significant differences between them.

Tang (2017) reported that labially coated stainless steel wires had significantly higher loading properties owing to greater overall cross section. This is agreeing with the findings of the present study regarding the properties of labially coated wires from DB.

Other studies **(Elayyan *et al.*, 2008; Elayyan *et al.*, 2010; Lijima *et al.*, 2012; Bradley *et al.*, 2014; Pop *et al.*, 2015)** tested NiTi wires and also concluded that coated wires showed lower loading and unloading values as compared to uncoated wires.

Abaas and Alhuwaizi (2015) reported that epoxy coated NiTi wires produce lower forces than polymer and Teflon coated wires. This comes in line with the findings of the present study.

The spring-back of GH, Highland, Dany, DB and Hubit wires were all significantly higher than the uncoated control wires for both wire dimensions which reflect their abilities to be bent to a large extent without permanent damage and store more energy before deactivation as compared to un-coated wires **(Kusy, 1997)**. These higher values could be attributed either to high values of yield strength or to low values of modulus of elasticity. The labially coated TP and RMO wires' readings were very comparable to the uncoated control wires for both wire dimensions since they have comparable yield strength and elastic modulus.

From these results, it seems that the change in cross section dimensions of wire did not directly affect the spring back property since it is the ratio of yield strength (MPa) to the modulus of elasticity (MPa).

In summary, the size of wire is the main but not the only influencer to the mechanical properties which may be changed with variation in the alloy composition or manufacturer processing; therefore, the manufacturer should improve the alloy mechanical properties when coating a smaller inner core, or preserve the underlying wire dimensions in order to maintain the mechanical properties.

4.3 Frictional resistance force of the wires

The present study revealed that friction increases with larger dimension coated stainless steel arch wires from 0.016''×0.022'' to 0.019''×0.025'' which agrees with many studies that studied uncoated and coated wires **(Ogata,1996; Al-makhzomi, 2013; Patil *et al*., 2016; Farronato *et al*., 2012; Shahabi, 2017).**

On the other hand, an in vitro simulation study **(Ireland, 1991)** found that smaller dimension wires lead to increased friction and attributed this to greater tipping of the teeth during sliding, however teeth tipping was not tested in the present study.

Moreover, other studies did not find a considerable relationship between dimension and friction **(Tidy, 1989)**. These findings disagree with the current study and the contradictory results are likely to be due to different experimental conditions since this study measure frictional force in simulated bodily tooth movement.

The maximum static frictional forces of the coated wires were lesser to or higher than those of the uncoated wires, and there was a variation in the degree of change related to multiple factors such as arch wire dimension, type and thickness of coating, hardness, surface roughness, and modulus of elasticity as it is reported by many studies (**Kumar, 2014; Choi, 2015; Muguruma, 2017**).

Although, some manufacturers try to coat the wires only on the labial surface to reduce surface roughness and minimize the effect of sliding mechanics, the maximum static frictional forces of labially coated wires from TP (polymer coated), DB (Teflon coated) and RMO (Teflon coated) were higher than uncoated wires. These values could be attributed to the greater width of the coated wires resulting from the additional coating layer or to higher elastic modulus.

These results come in agreement with **Mugurumaa (2017)** who concluded that friction of the coated stainless steel wires is affected by the total cross section

and inner core dimension and not by surface roughness and suggest that the high elastic modulus of wires may increase the wire binding at the edge of the bracket.

Despite the larger cross section of coated wires from Dany group which are coated with polymer layer all around the wire, their frictional resistance force was less than that of labial coated and uncoated wires. These reduced values may be the cause of thin thickness of coating layer or due to high spring back of these wires which resist binding with the edge of the bracket.

All fully coated arch wires show less static frictional forces than non-coated control group. **Jang *et al*., (2011)** concluded that the presence of coat has no effect on friction at 0˚ angle which is in line with our study. However, this disagree with the study performed by **Rudge *et al*. (2015)** which demonstrated that friction of fully coated wires was equal or higher than non-coated wires at all bracket wire angulation due to unknown reasons as stated by the researchers.

These findings may be attributed to the dimension of these wires which are less than the required stated dimensions. However, it should be noted that the friction of the coating layer differs when tested in vitro than in vivo after 4 to 6 weeks (**Rongo, 2014**).

To compare between the four fully coated wires, the highest values among both wire dimensions were with the polymer coated Dany wire, despite the polymer coating material having a smooth surface roughness as found by many studies that these polymer coated wires have a silver primer between the inner core and outer coating layer which improve cohesion between polymer and stainless steel, thus reducing surface roughness and friction **(Elayyan *et al.*, 2010; Burstone *et al.*, 2011; da Silva *et al.*, 2013; Neal, 2013)** but this high values could be strongly attributed to the greater height and width of Dany wires as compared to the other fully coated types.

It is observed that GH and Highland wires which are coated with Epoxy have nearly the same values with no significant differences between them. Their frictional resistance forces were more than Hubit wires owing to the relatively thick coating layer (0.35 to 0.65 mil for GH and Highland wires and 0.29 to 3.1 for Hubit wires).

Among the four types, the least values appeared with the Teflon coated wires from Hubit. This may be the results of thin coating layer of Teflon (0.29-3.1 mil) as compared to the epoxy coated wires (0.35-0.65 mil) or to physical characteristics of Teflon material such as surface roughness, hardness and elastic modulus which could reduce friction (**Muguruma, 2017**) however further researches are needed to improve this hypothesis.

Lower values of Teflon coated wires has been explained by many studies as the thin coating layer minimize the accumulation of Teflon material in the bracket slot which in turn reduce friction **(Kaphoor and Sundareswaran, 2012; Zegan *et al*., 2012)** and this is consistent with the present findings.

Krishnan *et al.* (2015) found that surface roughness of Teflon coated arch wire is less than other types including epoxy material which contribute to minimize resistant to sliding. This come in line and agrees with our results. However, **Doshi (2011)** concluded that there is no correlation between wire surface roughness and friction but the investigator measure different wire bracket combination and dose not measure coated arch wire**.**

The least values of friction for the Teflon coated wires of the present study is agreed with the study performed by **Agha (2014)** who attributed the result to the thin coating layer and support the study with **McCook (2004)** who concluded that Teflon material has a lower coefficient of friction when compared to epoxy material.

In this study, the differences in static frictional forces according to the measured dimensions of wires and to the type and thickness of coating layer were due to the direct influence of several other factors, so investigators should consider the physical characteristics of coating materials such as surface

roughness, corrosion, creep and relaxation, and the manufacturing processes (polishing, heating treatment, etc.), **(Ryu *et al*., 2015)**, bond strength between the coating layer and the inner core, elastic modulus of coating layer **(Mugurumaa *et al*, 2017).** All of these factors can affect resistance to sliding (friction).

All these results are susceptible to a great change in the oral cavity due to the effect of several functional factors such as chewing, swallowing, and speaking, as well as the tissues that are in contact with the appliance **(Kim, 2014; Prashant *et al*., 2015; Rongo, 2014; Pop *et al*., 2017).**

In the present study, a variation of the results between groups has been found among wire dimensions (0.016''×0.022'' and 0.019''×0.025'') which could be the result of variation in the coating thickness and dimension that differ among the same manufacturer without dependence on a stable standard as it is observed with the results of metallurgical microscope.

4.4 Strengths of the Study

A number of factors contribute to the strength of this study which are:

1- Comparisons are made between coated and un-coated arch wires from eight brands with two dimensions which is more than any current publication known to the authors.
2- Only a few studies (at the time of writing) are present for comparison between coated and un-coated wires.
3- Three-point bending test is performed in accordance with specification of the ISO standard 15841:2014 which is developed by American Dental Association as a result of the difficulty often encountered by clinicians in making meaningful comparisons between wires using the information currently available from manufacturers and suppliers while other studies use an arbitrary specification making comparison problems.
4- It is the first study that measure the wire size by a microscope in addition to the measurement of coating layer thickness and inner alloy dimension.

4.5 Limitation of the study

This study is limited by the lack of information from the manufacturer regarding the manufacturer processing and the method and temperature used during application of coating layer. Once again there is no standard in the manufacturing process related to the thickness of coating layer and arch wire dimension.

A true understanding of the behavior of wires is also limited due to lack of information available about the exact composition and ratios of elements of stainless steel wire and coating layer.

Comparison with previous studies is complicated and limited to some extent by a number of factors including:

1- Limited number of studies were performed between coated and un-coated wires.
2- Lack of a universal standard arch wires and lack of standard testing equipment.
3- Measurements of dimension by the metallurgical microscope is a new technique advised by the authors.

4.6 Clinical Consideration

1- The orthodontist must be aware that the manufacturers mostly state the dimension of coated arch wires including the thickness of coating layer and not the real inner core stainless steel dimension which greatly affect the mechanical properties of wires.
2- Arch wires with coating layer only on the labial surface nearly deliver the required force as the non-coated stainless steel wire.
3- A larger nominal wire dimension must be considered than would usually be selected when using full coated stainless steel wire.
4- The orthodontists must have the knowledge during selection of coated arch wires as reduced friction depends on many factors including wire dimension; but generally, Teflon coated wires would be appropriate with reduced friction treatment when using full coated wires.
5- Labial coated stainless steel wires in the present study have the advantage of increased friction which is useful in the finishing stage of clinical treatment.

Chapter Five

Conclusions & Suggestions

Chapter Five

Conclusions and Suggestions

5.1 Conclusions

Within the limitation of the study, we can conclude the following:

1- According to the method used, measured wire dimension does not match that stated by the manufacturer.

2- The decreased inner alloy dimension plays the main role in reduction of mechanical properties of wires; moreover, the alloy's composition and manufacturer processing.

3- Full coated arch wires show reduced mechanical properties as compared to their non-coated control and labial coated arch wires due to decreased inner alloy core dimension to compensate for the thickness of coating layer.

4- Mostly, labial coated stainless steel wires have nearly the same (comparable values) mechanical properties as their non-coated control arch wires.

5- The frictional forces of coated wires differ from uncoated control being higher in the labially coated wires and lesser in the fully Teflon coated wires owing to differences in the total wire dimension, inner core stainless steel dimension, modulus of elasticity, thickness of coating layer, and physical properties of coating materials.

5.2 suggestions for future studies

1- Study the mechanical properties of coated nickel titanium wires.
2- Using different coated stainless steel wires from different manufacturer.
3- Using tensile test in addition to three-point bending test and comparing between the two results.
4- Study the coated wires with scratch resistance and nanohardness test and analyzing the surface roughness of coating materials.
5- Conducting the study in vivo after 4-6 weeks of oral exposure to assess the effect of oral environment on the properties or coated wires.

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جمهورية العراق
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الخواص الميكانيكية لأسلاك تقويم الاسنان الفولاذية الغير قابلة للصدأ المغلفة (دراسة مختبرية)

رسالة مقدمة الى كلية طب الاسنان – جامعة بغداد
كجزء من متطلبات نيل درجة الماجستير
في تقويم الاسنان

من قبل
احمد عبد الحسين عباس
بكالوريوس طب وجراحة الفم والاسنان

بأشراف
أ.د. أكرم فيصل الحويزي
بكالوريوس طب وجراحة الفم والاسنان
ماجستير تقويم الاسنان
دكتوراه وقاية الاسنان

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الخلاصة

نظرا لزيادة الطلب على أجهزة تقويم اسنان ذو جمالية أفضل، اقبلت الشركات المختصة على تصنيع اسلاك تقويم الاسنان تجميلية والتي تحمل مواصفات جمالية إضافة الى خواص ميكانيكية جيدة.

صممت هذه الدراسة لقياس وتقييم الابعاد الحقيقية، سمك المادة المغلفة، والخواص الميكانيكية لأسلاك التقويم الفولاذية الغير قابلة للصدأ المغلفة بعدة أنواع من المواد المغلفة والتي تم تزويدها من سبع شركات وبحجمين مختلفين (''0.019×''0.025 and ''0.016×''0.022) بمجموع 238 سلك فولاذي مغلف و34 سلك غير فولاذي غير مغلف زودت كنماذج تحكم.

تم قياس الابعاد وسمك المادة المغلفة باستخدام جهاز (metallurgical light microscope) وتم قياس الخواص الميكانيك باستخدام جهاز (universal testing machine) وذلك باستخدام اختبار الانحناء ثلاثي النقاط لقياس الخواص الميكانيكية واختبار الاحتكاك لقياس قوة الاحتكاك وتم تحليل النتائج احصائيا باستخدام اختبار(ANOVA) و (LSD).

أظهرت نتائج الدراسة ان هناك اختلاف نسبي بين ابعاد الاسلاك المقاسة نسبة الى الابعاد المذكورة من قبل الشركة وكذلك اختلاف في سمك المادة المغلفة. اسلاك التقويم من شركة (DB) وشركة (Dany) أظهرت اعلى قيم لقوة الانحراف القصوى، مقاومة الخضوع ومعامل المرونة. اسلاك التقويم المغلفة من شركة (TP) و (RMO) أظهرت خواص ميكانيكية مقاربة لخواص الاسلاك الغير مغلفة بينما الاسلاك من شركة (GH) و (Highland) و (Hubit) أظهرت اقل قيم للخواص الميكانيكية. اختبار قوة الاحتكاك اظهر اعلى نتيجة لأسلاك التقويم المغلفة من الجهة الامامية لشركات (DB) و (TP) و (RMO) بينما اقل قيم ظهرت مع الاسلاك المغلفة من جميع الجهات من شركة (Hubit) بسبب اختلاف ابعاد الإسلاك وسمك المادة المغلفة إضافة الى معامل المرونة وخشونة سطوح الاسلاك التقويمية.

مما تقدم أظهرت نتائج هذه الدراسة ان ابعاد السبيكة الداخلية لأسلاك التقويم وسمك المادة المغلفة يلعب دور أساسي في تحديد الخواص الميكانيكية لأسلاك التقويم مغلفة بمواد تجميلية.