# EFFECT OF AGEING TREATMENT ON FORCE-DEFLECTION BEHAVIOUR OF NiTi WIRE IN THREE-POINT AND THREE-BRACKET SYSTEM


By:

**NURUL SYIFA' BINTI SHAMSUL KHAIRI**

(Matrix no: 143769)

Supervisor:

**Dr. Muhammad Fauzinizam Razali**


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Engineering Campus
Universiti Sains Malaysia

## DECLARATION

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This thesis is the result of my own investigation, except where otherwise stated. Other sources are acknowledged by giving explicit references. Bibliography/references are appended.

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## LIST OF ABBREVIATIONS

| | |
|---|---|
| NiTi | Nickel Titanium |
| UTM | Universal Testing Machine |
| TTR | Temperature Transition Range |
| SME | Shape Memory Effect |
| CoCrNi | Cobalt Nickel Chromium |
| ßTi | Beta Titanium |
| SMA | Shape Memory Alloy |
| SIM | Stress Induced Martensite |
| IBD | Interbracket Distance |
| DSC | Differential Scanning Calorimetry |
| OTM | Optimal Tooth Movement |

**KESAN PENYEPUHLINDAPAN TERHADAP TINGKAH LAKU DAYA-LENTURAN OLEH DAWAI ARKUS NiTi DI DALAM SISTEM TIGA TITIK DAN TIGA PENDAKAP**

## ABSTRAK

Dawai arkus NiTi yang digunakan untuk meratakan dan menjajarkan gigi diperlukan untuk mempunyai julat operasi yang baik yang boleh memberikan daya yang ringan dan berterusan kepada gigi. Semasa fasa meratakan gigi ketika rawatan ortodontik, dawai arkus superelastik NiTi yang bersaiz segi empat tepat kebiasaannya digunakan kerana keupayaannya untuk menghasilkan tork dan menyelaraskan kedudukan gigi. Namun begitu, dawai arkus segi empat tepat hanya boleh digunakan selepas dawai bersaiz bulat telah meratakan gigi ke tahap tertentu. Ini disebabkan oleh kekakuan dan tork yang tinggi yang dihasilkan oleh dawai segi empat tepat. Kajian ini direka bentuk untuk menyiasat kesan penyepuhlindapan terhadap tingkah laku daya-lenturan dawai superelastik NiTi yang dilenturkan dalam tetapan tiga titik dan berbeza susunan pendakap gigi. Dalam kes ini, lenturan tiga titik dijalankan terlebih dahulu untuk mengukur ubah bentuk lenturan bagi 0.016 × 0.022 inci dawai arkus superelastik NiTi segi empat tepat yang dilenturkan selepas penyepuhlindap pada pelbagai suhu dengan tempoh yang berbeza. Selepas itu, lenturan dalam sistem tiga pendakap dilakukan menggunakan dawai yang telah dipenyepuhlindap dengan daya pemunggahan terendah untuk setiap tempoh penyepuhlindapan yang dipilih berdasarkan data daripada ujian lenturan tiga titik. Penemuan kajian mendedahkan bahawa daya lentur yang dijana oleh dawai arkus NiTi semasa kitaran pemunggahan telah diturunkan dengan berkesan oleh penyepuhlindapan. Cerun dataran tinggi, yang menggambarkan ketekalan daya dawai, berbeza dengan ketara antara dawai yang sedia ada dan yang telah dipenyepuhlindap. Semasa lenturan tiga titik, dawai penyepuhlindapan pada 490 °C selama 45 minit memberikan daya pemunggahan yang paling rendah (1.72 N) manakala dawai sedia ada menghasilkan 2.99 N. Dalam system tiga pendakap, nilai daya terkecil dihasilkan oleh dawai penyepuhlindapan pada 490 °C selama 45 minit dengan pengurangan daya sebanyak 11 hingga 33% berbanding dengan dawai sedia ada. Parameter penyepuhlindapan dengan penghantaran daya tetap terbaik dalam lenturan tiga pendakap dihasilkan pada 460 °C selama 30 minit. Ia juga diperhatikan bahawa dataran daya rata telah dihasilkan semasa pemunggahan apabila dawai arkus digunakan dalam

jarak antara pendakap yang lebar, iaitu 8.5 dan 9.0 mm. Penyepuhlindapan boleh digunakan untuk melaraskan lagi tingkah laku ubah bentuk dawai arkus superelastik NiTi supaya ia akan menggunakan jumlah daya yang sesuai semasa rawatan meratakan gigi.

# EFFECT OF AGEING TREATMENT ON FORCE-DEFLECTION BEHAVIOUR OF NiTi WIRE IN THREE-POINT AND THREE-BRACKET SYSTEM

## ABSTRACT

NiTi archwires used for levelling and alignment of the teeth are required to have a good operating range that can provide light and constant force applied to the teeth. During the teeth levelling phase of orthodontic treatment, a rectangular superelastic NiTi archwire is usually used due to its ability to transmit torque and align the position of tooth. However, rectangular archwire can only be applied after round wire has levelled the teeth to a certain extent. This is due to the high stiffness and high torque that is produced by rectangular wire. This study was designed to investigate the effect of ageing treatment on the force-deflection behaviour of superelastic NiTi wire bent in three-point and different interbracket settings. In this case, three-point bending was first conducted to measure the bending deformation of 0.016 × 0.022-inch of rectangular superelastic NiTi archwires bent after being aged at various temperatures with different durations. Subsequently, bending in a three-bracket system was performed using aged wires with the lowest unloading force for each ageing duration, which were selected based on the data from three-point bending test. Findings of the study revealed that the bending force generated by the NiTi archwire during the unloading cycle was effectively lowered by the ageing treatment. The plateau slope, which depicts the force constancy of the wire, considerably differed between as-received and aged wires. During three-point bending, ageing setting of 490 °C at 45 minutes exerted the lowest unloading forces (1.72 N) whereas as-received wire produced 2.99 N. In the three-bracket system, the least force magnitude was yielded by wires aged at 490 °C for 45 minutes with 11 to 33% force reduction as compared to as-received wires. Ageing parameter with the best constant force delivery in three-bracket bending is produced at 460 °C for 35 minutes. It was also observed that flat force plateaus were produced during the unloading when archwires were employed in wide interbracket distance, which are 8.5 and 9.0 mm. The ageing treatment approach may be utilised to further adjust the superelastic NiTi arch wire's deformation behaviour so that it will exert the appropriate amount of force during tooth levelling treatment.

# CHAPTER 1

# INTRODUCTION

## 1.1 Research Background

The majority of malocclusion patients are prescribed orthodontic levelling treatment to correct the vertical discrepancy between their teeth. Typically, fixed appliance therapy also known as braces is used to treat this condition, in which an archwire is positioned inside the bracket slots while following the irregularity of bracket position. This treatment may take from 14 to 33 months depending on the condition of the teeth [1]. However, a longer treatment time is expected for poorer malocclusion cases. Therefore, the application of a system that is properly supported by brackets, archwires, elastic ligatures, and other accessories, as shown in Figure 1.1, is required for optimal control of tooth movement. This treatment helps to improve the dental alignment by shifting the position of the teeth. The function of bracket is to place the archwire in the bracket slot, which will allow a controlled movement of the teeth whereas elastic ligature is generally used to hold the archwire in the bracket slot.

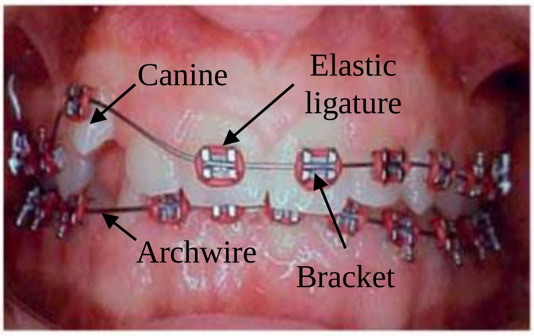


Figure 1.1: NiTi wire canine traction [2]

Archwire plays a significant role in tooth movement. The force applied by archwire should be ideal to cause the least amount of discomfort to the patient, as well as to reduce the risks of root resorption [3] and tissue hyalinization [4]. The archwire should behave elastically when it is subjected to a force for a period of time, which can last for weeks to months. Having said that, superelastic NiTi wire is the most commonly used wires for teeth alignment as it produces less painful bone remodelling [5]. This is due to its mechanical properties that promote light, continuous force across a broad range of wire activation.

Although several studies have been conducted related to force-deflection behaviour of NiTi archwires and frictional force at wire-bracket interfaces, most of the studies mainly focused on the three-point bending test, which disregarded the function of bracket engagement. Hence, the result obtained from the experiment is deemed unreliable for use in oral clinical settings [6]. Without the bracket configuration, it eliminates the presence of frictional force at the contact region of NiTi wire and bracket corner during the test. Montasser et al. [7] found that the loss of applied force can range from 12% to more than 70% due to frictional force. That being said, greater mechanical forces are required to be applied to overcome the friction.

Various studies have evaluated the force-deflection behaviour of superelastic NiTi archwire in different bracket configurations. For example, an experimental study was conducted to evaluate the bending deformation behaviour and the effect of contact friction towards bending deformation of NiTi archwire [8]. Interbracket distance of 7.5 mm were implemented in the study as shown in Figure 1.2(a). This arrangement represents the canine, first and second premolars of the maxillary jaw. On the other hand, Figure 1.2(b) shows a three-point bending setup with a NiTi wire. The major difference between these tests is the presence of orthodontic brackets, which is very important to study the force-deflection behaviour of the archwire.

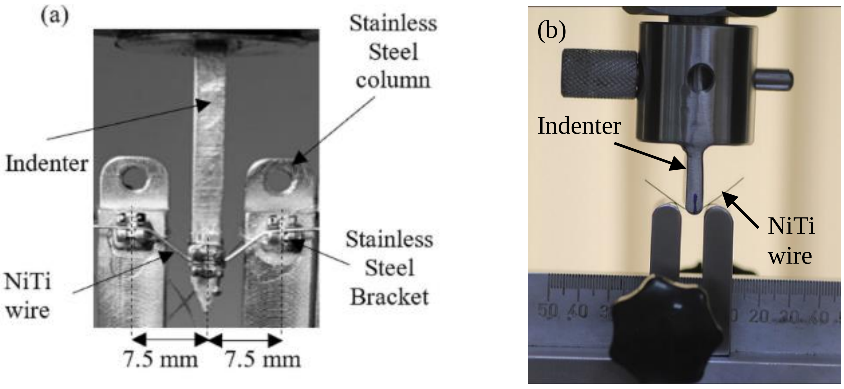


Figure 1.2: (a) Three-bracket bending setup with stainless steel bracket [8] and (b) Three-point bending setup [9]

Figure 1.3 shows the load-deflection curves of NiTi archwires in three-point and three-bracket models. There is a substantial difference that can be seen on the loading curves of both models. In the three-point system, the archwire shows loading and unloading curves over a force plateau, which indicates that the deformation of archwire began under superelastic behaviour. In the presence of brackets, however, these loading and unloading curves show positive and negative gradient slopes, respectively. This is due to the wire curvature pressing hard against the bracket corner at large deflection, resulting in a gradual increase in friction intensity [10]. Subsequently, the high friction caused at the start of the bending recovery considerably decreased the unloading force, which then gradually increased as the wire deflection was reduced.

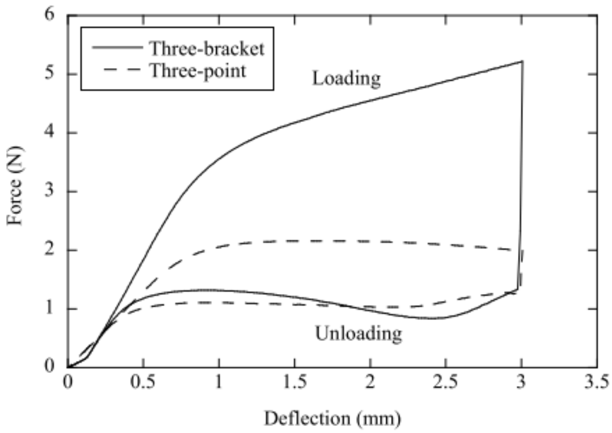


Figure 1.3: Load-deflection curves of NiTi archwires in three-point and three-bracket bending [10]

Heat treatment, such as ageing, is an excellent approach for increasing the yield strength and improving the mechanical properties of NiTi shape memory alloy due to the formation of Ni-rich precipitates [11], [12]. In general, $Ni_4Ti_3$ precipitates can form in Ni-rich NiTi samples that have been aged at a specific temperature. This formation not only modifies the mechanical behaviour, but also functions as a hardening agent inside the matrix of austenite, B2 providing strength to the alloy [13]. Nonetheless, coherent $Ni_4Ti_3$ precipitates are only effective for enhancing the alloy's shape memory behaviour at a certain temperature and ageing time [12]. According to [12], the sample that was aged at 450 °C produced fine and dense $Ni_4Ti_3$ precipitates.

However, to our knowledge, there have not been many studies conducted on the effect of ageing treatment to improve the bending behaviour of the orthodontic archwire as most ageing studies stop at observing the tensile behaviour of NiTi alloy. In other words, no one expands the work to bending in bracket system. Therefore, this project aims to produce highly flexible rectangular NiTi archwire, which can be done by ageing treatment.

In this study, an experimental test was conducted to study the effect of ageing treatment on the force-deflection behaviour of NiTi wire in the three-point and three-bracket system. Ageing treatment was performed before proceeding with the tests. The force-deflection curves are presented as the findings of the project. This approach may help the orthodontists in identifying the most appropriate wire-bracket combination and condition to produce constant force delivery in order to achieve optimal tooth movement.

## 1.2 Objectives

1. To identify the suitable heat treatment parameters for lowering the unloading force of NiTi archwires.
2. To study the force-deflection behaviour of aged NiTi archwire in different dental interbracket settings.

## 1.3 Problem Statement

Friction at the contact region of NiTi wire and bracket corner is a major impediment to tooth movement as it changes the constant force behaviour of superelastic NiTi wire into gradient behaviour. The gradual increase in the frictional resistance due to wire bending stiffness and amount of deflection has influenced the formation of the gradient bending force pattern. This is due to the bracket corner being hardly pressed by the wire curvature at great deflection. Consequently, frequent force changes over wire deflection may cause tooth movement to be delayed. Thus, a constant force is required to induce an optimal tooth movement rate. Ageing treatment can be used to reduce the force slope by lowering the stiffness of rectangular NiTi wire.

## 1.4 Scope of Project

In this study, an experimental test was developed to identify the suitable heat treatment parameters for the production of NiTi archwire. Ageing treatment with multiple ageing parameters was carried out beforehand to improve the material characteristics of the NiTi wires. The most appropriate ageing parameters were selected to undergo three-point and three-bracket bending tests. The three-point bending test was conducted at 36 °C to observe the force-deflection behaviour. Bending in bracket system was carried out after three-point bending test using different interbracket distances at 36 °C to evaluate the true flexural behaviours of the archwire.

# CHAPTER 2

# LITERATURE REVIEW

## 2.1 Introduction

This chapter focuses on the review of previous research that is related to this study.

## 2.2 Orthodontic Treatment Stage

Orthodontic treatment is a method of improving the appearance of the teeth by straightening or moving them into a better position. The length of treatment typically takes from 6 to 30 months to complete. Based on a randomized clinical trial, the mean treatment duration for the 0.018-inch and 0.022-inch bracket slot groups was 29.3 months and 31.2 months, respectively [14]. The treatment duration is primarily determined by the classification of the malocclusion, the age of the patient, and the type of dental devices used.

There are three major stages of the treatment, which are aligning and levelling stage, correcting the bite stage, and the finishing stage. During the initial stage, which is the aligning and levelling stage, a constant and light force will be applied along the archwire attached to the brackets, causing compression and tension on the teeth and periodontal ligaments. This force will assist in aligning the teeth in a relatively short time. In certain cases, the results are noticeable in less than 3 months depending on the initial condition of the teeth. Clinical studies revealed that the optimal tooth movement (OTM) rates range between 0.55 and 2.44 mm each month during the initial alignment [15].

The second stage focuses on correcting the bite of the top and bottom teeth and this may take anywhere between 12 and 18 months to complete. During this stage, fixed appliances are typically applied to both the maxillary (upper) and mandibular (lower) teeth so that the upper teeth fit perfectly over the lower teeth. Usually, a stiffer and wider archwire is used in the bite adjustment stage.

The final stage aims to reduce the gap between the teeth. Finishing is usually the last step before discontinuing active treatment. The function of finishing is to ensure that the teeth and related structures are positioned appropriately so that the teeth achieve better stability, improve the appearance, optimise the stomatognathic system (a complex sensory system that performs functions such as yawning, swallowing, breathing, speech, mimicking, and chewing [16]) functions, and enhance periodontal (gum) health [17].

Orthodontic tooth movement is greatly influenced by the amount of force applied. According to Nakano [18], the lightest force caused the most tooth movement, while heavier forces caused less tooth movement in the tipping and bodily tooth movement. When the force magnitude of either tipping or bodily tooth movement surpassed a certain limit, the amount of tooth movement reduced. Forces of 0.1–0.6 N are considered ideal for vertical tooth movements, depending on the type of tooth [19] and this was consistent with the findings of Proffit (0.34−0.59 N) [20]. In addition, forces ranging from 0.5 to 1.0 N also seem most appropriate to facilitate tooth movement with potentially less side effects as shown in Figure 2.1 [21], [22]. Besides, the study discovered that there will be an increase in the rate of orthodontic tooth movement if forces greater than 2.5 N are applied but there are also negative side effects, such as severe pain or orthodontically induced external root resorption.

| Group | Force cN | OTM mm/week(range) |
|---|---|---|
| Low | <100 | 0.23-0.44 |
| Moderate | 100-150 | 0.16-0.47 |
| High | 150-250 | 0.1-0.46 |
| Very high | 250-400 | 0.34-0.49 |

Figure 2.1: Rate of orthodontic tooth movement (OTM) per week [22] (Note: 1 cN = 0.01 N)

## 2.3 Type and Size of Archwire and Dental Bracket

In orthodontics, an archwire is a wire that conforms to the dental arch or alveolar, which is mainly used to correct the irregularities in teeth position with dental braces by exerting a gentle force. The archwire connecting the braces must be able to revert to its original shape if it is deformed or bent. As the teeth straighten, the wire will be replaced with a stronger, stiffer, and less elastic wire. The properties and behaviour of the archwire are determined based on the type and size of the wire.

NiTi alloy, which consists of 55% nickel and 45% titanium, is frequently used during the initial stage of treatment; i.e. levelling and alignment. It has low stiffness and high springback, NiTi archwire is commonly used in the early stage of orthodontic treatment as it is needed to align crooked teeth with gentle force. The properties of NiTi wire are widely known as shape-memory alloy and superelasticity.

### 2.3.1 Shape Memory Effect (SME) and Superelastic NiTi Archwire

Shape memory effect allows the alloy to revert to its previous shape after undergoing massive inelastic deformation through heating. This type of alloy promotes excellent deformation recovery, with the ability to regain its original length ten times longer than traditional alloys [23]. Typically, NiTi shape memory alloy undergoes heat treatment to alter the deformation and mechanical behaviour to match the desired application and it exists when it is heat activated above the Temperature Transition Range (TTR) which ranges from 20.39 °C to 45.42 °C [24].

It is shown in Figure 2.2 that thermally induced SME follows the path A-B-C-D-A, which indicates a shape restoration of NiTi material that occurs through the transformation of deformed martensite to austenite phase [25]. The operating temperature for thermal shape memory should be lower than the Maternsitic finish ($M_f$) temperature. When martensite is subjected to an external load, it accommodates detwinning via a large inelastic strain at a constant stress plateau. The small elastic portion of detwinned martensite is recovered when the load is removed at B, leaving a significant amount of residual strain at C. Heating the alloy above Austenite finish ($A_f$) temperature at D caused the residual strain to gradually decrease as the alloy remembered its high-temperature structure. This shape memory effect is only applicable once, and the alloy must be deformed again to obtain the next cycle of shape recovery [23].

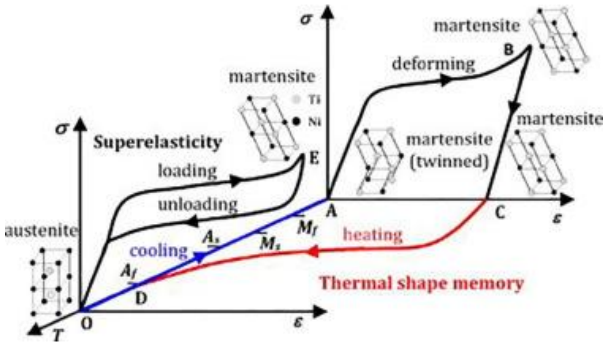


Figure 2.2: Thermal shape memory and superelasticity of NiTi [25]

The next properties of NiTi is superelasticity, which is defined as the immediate recovery of deformed shape memory alloys without residual deformation under large strain (6–13.5%) [26] when applied force is removed. In Figure 2.2, it is shown that superelasticity follows the path O-E-O. Superelastic wire does not necessitate any external media to regain its shape. In Figure 2.3, superelastic behaviour can be observed when the material temperature is higher than the Austenite start ($A_s$) temperature. However, for complete recovery with minimal residual deformation, the material temperature should be higher than the $A_f$ temperature but lower than the Martensite desist ($M_d$) temperature [26].

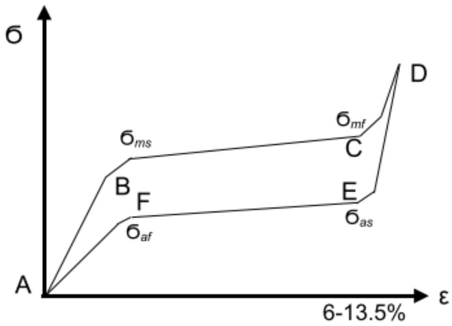


Figure 2.3: Stress-strain curve of superelastic wire [26]

Figure 2.4 shows the load-deflection curves of a pseudoelastic NiTi alloy and other conventional elastic material (i.e. stainless steel, CoCrNi, and ßTi) from a three-point bending test. It is observed that NiTi alloy produces the lowest force plateau with the force magnitude of 6.6 N among the others at the highest maximum deflection. The unloading force at the deflection of 2 mm is approximately 2.9 N. In addition, NiTi curve returned to the starting zero point, which indicates superelasticity.

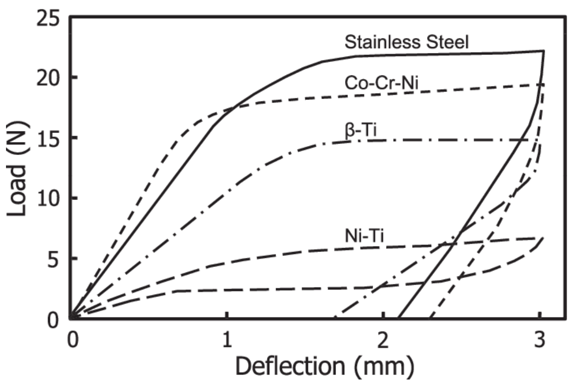


Figure 2.4: Load-deflection curve of pseudoelastic NiTi, stainless steel, CoCrNi, and ßTi [27]

Figure 2.5 compares the mechanical properties of a pseudoelastic NiTi alloy to a conventional elastic material. The excessive force zone represents the force that causes tissue damage, while the optimal force zone, ranging from 0.5 N to 2.5 N [22], demonstrates the effective force for bone remodelling. The effective strain range ($\varepsilon_{eff}$) is the range of activation available while moving the teeth. In comparison to stainless steel, NiTi alloy demonstrates a wide effective strain range over the optimal force zone. As a result, NiTi alloy has a wider activation range and requires less archwire adjustment to move the teeth to the final position.

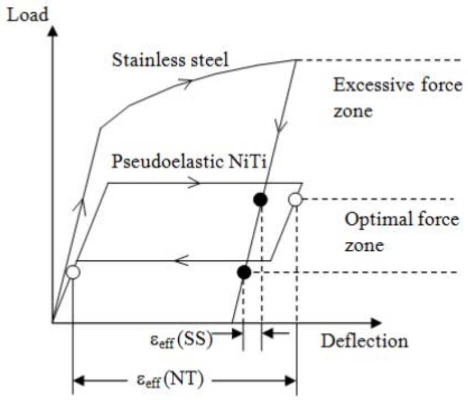


Figure 2.5: Load-deflection curve of pseudoelastic NiTi archwire and stainless steel [23]

### 2.3.2 Size of Archwire

Considering archwires are made of the same material, the smaller the size as in the cross-section of the wire, the higher the elasticity. There are three types of cross-section for orthodontic wires, which are round, square, and rectangular. Round wires are commonly used in the early stage of treatment to align the teeth as it is more elastic and have minimal friction compared to square and rectangular wires. Despite that, round wire has no torque control. Therefore, as the teeth become straighter, orthodontists will usually change the wire to rectangular wire.

It is unusual for the orthodontist to use rectangular wire at the beginning of treatment as it produces the most friction and torque. This may cause the wire to exert too much pressure on the bracket, causing it to break off from the tooth and cause discomfort to the patient. This assertion is validated by the earlier study in [28], where such huge difference in force magnitude between rectangular and round wires are shown in Figure 2.6. 0.018 × 0.025-inch rectangular wire, the highest size of rectangular wire, exhibits the highest force plateau magnitude and 0.016-inch round wire, which is the smallest diameter of round wire in the study, produces the lowest force plateau. It can be concluded that the wire stiffness is affected by the size of the archwire due to the moment of inertia of the wire geometry. Moreover, wire stiffness influences the bending moment and degree of deflection of the wire. Thus, it would be beneficial to the orthodontic field if the flexural properties of rectangular wire can be altered so it can be used in the early stage of treatment. It would be less time consuming and low cost if this objective can be achieved.

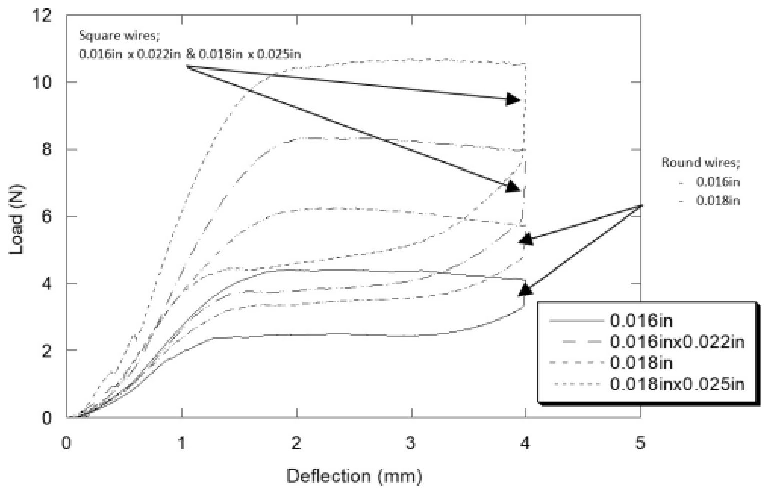


Figure 2.6: Load-deflection curve of rectangular and round NiTi wires in three-bracket system [28]

### 2.3.3 Type and Size of Bracket

Dental brackets are small squares that are bonded directly to the front of each tooth. They are available in various sizes and styles, which include metal, ceramic, and self-ligating. The function of bracket is to hold the archwire that has been discussed earlier.

In spite of the near universal use of brackets in orthodontic treatment, only few scientific studies have been conducted on the tolerances of bracket slots, which are especially important when using a preadjusted appliance. Orthodontists tend to rely more on peer advice and expert opinions, as  scientific data considered valuable only 42% of the time when choosing dental products. It is stated in [29] that oversized slots undermine the entire concept of preadjusted edgewise, which is initially designed to reduce wire bending.

Edward Angle [30] proposed the 0.022-inch by 0.028-inch bracket slot size in 1925, which allowed greater control of crown and root position with the metal archwires available at that time. However, the advancement of technology makes it possible to produce a smaller bracket slot size, and as a result, 0.018-inch bracket slot was successfully manufactured [30], [31]. Despite that, 0.022-inch bracket slot is still being used in clinical practice. Figure 2.7 displays the position of 0.018-inch archwire in the bracket slot of 0.022 inch and 0.018 inch. It can be seen that the 0.018 inch archwire is loose in the bracket slot of 0.022 inch, however it is slightly tight in the bracket slot of 0.018 inch.

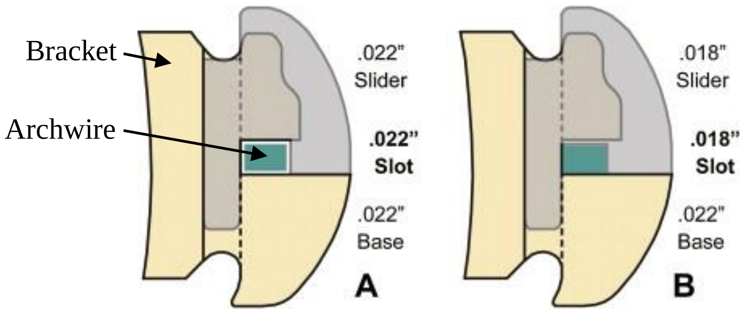


Figure 2.7: 0.018-inch archwire in the bracket slot of (a) 0.022 inch and (b) 0.018 inch [32]

With 0.022-inch bracket slot, it has been suggested that overbite reduction and closure of any residual extraction space could be more efficient. This is due to the space between the working archwire (0.019 by 0.025-inch stainless steel) and the 0.022-inch bracket slot that allows the bending of exaggerated bite-opening while still being able to be fitted with minimal effort [30]. Moreover, 0.022-inch bracket slot allows wires more freedom of movement during the early alignment stage, resulting in lighter forces.

Based on a systematic review [31], the 0.018-inch bracket slot size was reported to be more efficient than the 0.022-inch counterpart in three of the four studies. One of the studies discovered that patients treated with 0.022-inch bracket slots had a 3.8-

month increase in treatment duration [33]. Furthermore, in another study [34], the treatment with 0.022-inch bracket slots was found to be 9.5 months longer on average than treatment with 0.018-inch bracket slots. Nevertheless, there are no data to support the superiority of one system over the other because the risk of bias inherent in the design of the analysed studies prevented a definitive conclusions.

## 2.4 Superelastic Behaviour of NiTi Alloy under Tensile, Three-Point and Three-Bracket System

### 2.4.1 Tensile Test

The behaviour of shape memory alloy wires in restoring their original form can be divided into two categories, which are superelasticity (pseudoelasticity) and shape memory effect. NiTi superelastic shape memory alloy (SMA) wire is commonly used in orthodontic treatments with a maximum strain of 6% to 13.5% [26], making it one of the best shape recovery limits in alloy materials. This pseudoelastic SMA material can recover its original shape in tensile configuration with up to 8% strain [35]. This material exhibits a loading plateau after exhibiting a first linear elastic behaviour. Figure 2.8 shows stress–strain curves of a superelastic NiTi wire. From a microstructural standpoint, this superelastic behaviour occurs when its austenitic crystal structure transforms to a martensitic structure (stress induced martensite (SIM)), which begins as the wire is stressed further at larger deflection, transforming the linear curve on a force-deflection plot into a plateau [35], [36]. The superelastic material displays a second stress plateau during unloading, which represents the unloading constant force generated by the archwire against the resistance of the teeth. The microstructural state shifts from stress-induced martensite to austenite during this step.

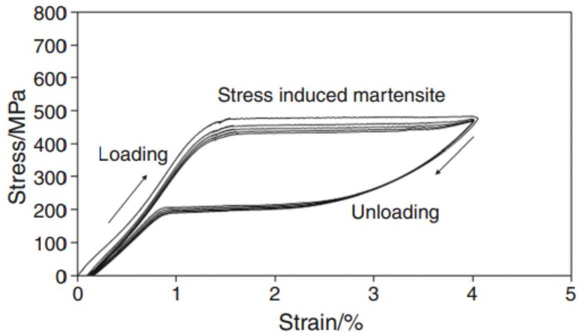


Figure 2.8: Stress-strain curves of superelastic NiTi wire during five tensile cycles at T > $A_f$ [35]

According to Hassan et al. [26], NiTi archwire with the austenite finish temperature of 20 °C and -15 °C has shown excellent recovery to its original form for tensile test up to 6% strain compared to the archwire with austenite finish temperature of 45 °C. In addition, less than 50% – 60% stress is required to complete martensite transformation for NiTi archwire with a diameter of 0.02 inch in comparison with 0.03 inch-diameter wire, which required higher stress. Hence, a higher force is needed for NiTi archwire with higher austenite finish temperature to activate the martensite transformation.

### 2.4.2 Three-Point Bending

In a three-point model, superelastic NiTi wire exhibits bending force over a force plateau when loading and unloading [37]. In reality, this constant force behaviour is a portion of interest as it signifies the ability of NiTi archwire to provide a constant and light force to the teeth. The force-deflection curves generated from three-point bending tests were considered to be inaccurate as the traditional bending model neglected the function of bracket engagement, consequently ignored the presence of friction between the wire and brackets. As a result, the bending deformation behaviour has a flat force plateau during both activation and deactivation [8], [38].

In a study done by Khatri et al. [4], they observed that 0.016-inch superelastic NiTi wire exerted a relatively high loading and unloading force at 37 °C compared to other materials, which are 2.34 N and 1.47 N, respectively. In addition, NiTi archwires had the greatest mean unloading force at the deflection of 0.5 mm, which is 0.75 N and second greatest at the deflection of 1.0 mm, 1.5 mm, and 2.0 mm, which are 0.97 N, 1.29 N, and 2.87 N, respectively. A study in [38] discovered that the unloading forces of 0.013-inch and 0.016-inch wire sizes of all tested NiTi wires were less than 1.47 N and it was found in a previous study that a force of 1.47 N produced maximum bending of the periodontium of a canine tooth during and after distalization.

Further study [39] of three-point bending was conducted at 37 °C with conventional 0.018-inch NiTi wires from different manufacturers as the wire parameters. As shown in Figure 2.9, all NiTi wires exhibited superelastic behaviour with hysteresis, which happens when the strain energy is lost as heat between loading and unloading due to internal friction in the material, and higher loading forces than unloading forces. Nonetheless, at the deflection of 1 mm and 2 mm, the curves differed in shape and delivered slightly different forces. Furthermore, fluctuations at the deflection of 0.5 mm on the unloading branch were drastic, as the majority of the NiTi wires had already left the unloading plateau or had a slight permanent (martensitic) deflection.

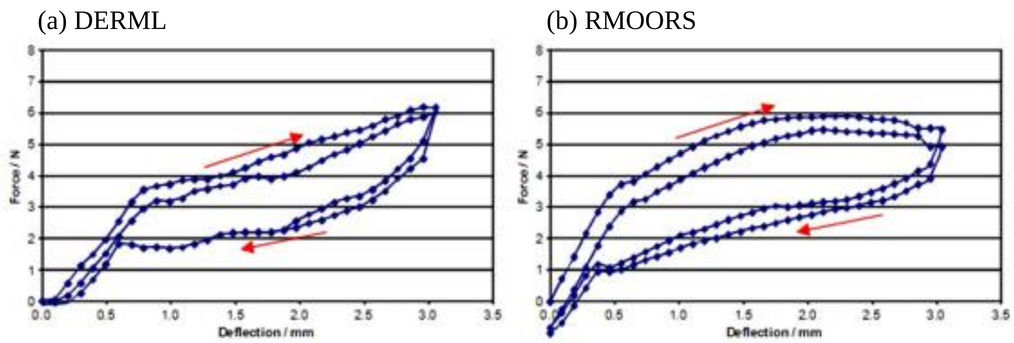


Figure 2.9: Force-deflection curve in three-point bending of conventional NiTi archwires from different manufacturers; i.e. (a) DERML (b) RMOORS [39]. Each sample was loaded twice, hence two loading/unloading curves are shown for each wire.

### 2.4.3 Three-Bracket System

Up to now, a variety of in-vitro bending models have been used to evaluate the flexural behaviour of NiTi archwires, including cantilever bending, three-point bending, and three-bracket bending. The latter bending system, which simulated the constraint bending condition of NiTi wires in the bracket slot, is identified to provide an accurate representation of wire deformation intraorally. When considering the dental bracket, the classic unloading plateau of NiTi archwire was reported to transform into a slope trend as soon as the wire recovered from large deflections (3.0 mm and above) [40]. The slope force plateau may be due to variations in the intensity of the frictional force as the wire deflected further at large deflection [41].

Figure 2.10 depicts force-deflection curves of conventional 0.018-inch NiTi wires from different manufacturers that was conducted at 37 °C in three-bracket system with elastic ligature. The interbracket distance used was approximately 8.5 mm with 0.022-inch bracket slot. As shown in the diagram, the forces are slightly greater than in three-point bending (see Figure 2.9) and the hysteresis curves of the NiTi archwires are wider. According to the study, this occurred due to the friction effect in the system as friction slightly increases the force when loading by preventing free sliding through the bracket slots [39], [42]. The loading forces were not discussed in the study. However, based on the figures, the positive slope represents the increase in force required to bend the wire as a result of the binding induced at the wire-bracket contact [36].

The average maximum loading forces of the NiTi wires from DERML and RMOORS are approximately 9.4 N and 8.6 N, respectively. Besides, the force is reduced during unloading because the wires do not slide back freely and the pressure on the loading tip is reduced. The forces on the unloading path for DERML NiTi wire at 1 and 2-mm deflections are 3.8 N and 3.4 N, respectively whereas for RMOORS, the wire produced 3.15 N and 2.4 N at the deflections of 1 and 2 mm, respectively.

(a) DERML (b) RMOORS

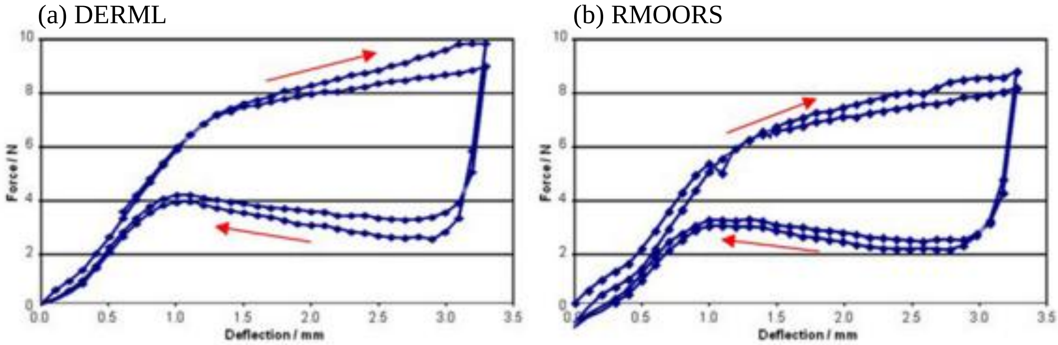


Figure 2.10: Force-deflection curves in three-bracket system of conventional NiTi archwires from different manufacturers; i.e. (a) DERML (b) RMOORS [39]. Each sample was loaded twice, hence two loading/unloading curves are shown for each wire.

In another study [8], the bending deformation behaviour of 0.016-inch NiTi wire had shown a slightly steep force plateau when it was used in 0.022-inch bracket slot stainless steel three-bracket bending configurations with the interbracket distance of 7.5 mm. This indicated the presence of greater friction at the contact region between the archwire and bracket corner, which was due to the smaller space between the brackets. In addition, based on the study, increasing deflection magnitude causes the activation force of the wire to continuously increase as well.

It is also mentioned that the force plateau gradient of the archwire showed a negative trend during deactivation, which causes a reverse phase transformation to begin at a lower force. This indicates that the spring back force generated by the wire during recovery was used to overcome friction at the brackets [36], [41], delaying the phase transition to a lower force level. The radius curvature of the wire decreased as the wire deflection recovered, hence reducing the normal force acting on the brackets as well as the friction. As the deflection recovered, the deactivation force increased indicating that the binding is weakening [6]. On top of that, it is interesting to mention that the amount of unloading force is most important since it is required to determine the magnitude of force delivered to the bracket [43]. Therefore, it is best to reduce the slope gradient to achieve a shorter treatment period [44].

## 2.5 Factors Influencing Friction during Archwire Sliding in Dental Bracket

The magnitude of the bending forces applied by the NiTi wire is known to be considerably affected by the magnitude of friction produced at the wire-bracket interfaces [10]. There are two basic types of friction, which are static friction and kinetic friction.

In orthodontics, static friction is more important than kinetic friction as tooth movement is not continuous and static friction needs to be overcome to initiate the tooth movement. Higher friction levels during sliding mechanics necessitate the use of greater orthodontic forces, which may limit the amount of orthodontic tooth movement achieved and disturb the anchorage control [45]. Stated in [46] that the percentage of applied force lost due to sliding resistance between the archwire and bracket ranges from 12% to 60%, which results in a decrease of tooth movement. This friction is the result of the binding force at the tip of the bracket [28]. In consequence, more force is required to achieve the desired result and eventually causes intense pain, anchorage loss, and root resorption. Hence, the frictional force should be controlled throughout the orthodontic treatment to allow optimal tooth movement. Several factors influence the frictional resistance produced during sliding mechanics, which includes archwire dimension and bracket slot size.

### 2.5.1 Archwire Dimension

Archwire physical and mechanical properties are considered as the major contribution in influencing the friction during treatment time. According to Dudani et al. [47], larger wire results in higher frictional resistance as there is very little free space in the bracket slot, thus the critical angle is met with less bracket tipping. Thus, it has been suggested that bigger archwire sizes should be avoided during the space closure stage [48].

Based on a sliding test conducted by Razali on 0.016 × 0.022-inch NiTi wire with 0.022-inch stainless-steel bracket slot [36], the maximum peak of static friction was observed at the beginning of the wire movement, followed by peaks representing kinetic friction that fluctuated at slightly lower values (see Figure 2.11). However, the true measurement of sliding friction is observed at the fluctuation of peaks. Based on the result of the study with the maximum peak force between between 5.32 N and 5.81 N, the coefficient value is within the normal range for a NiTi archwire and stainless-steel bracket combination [49].

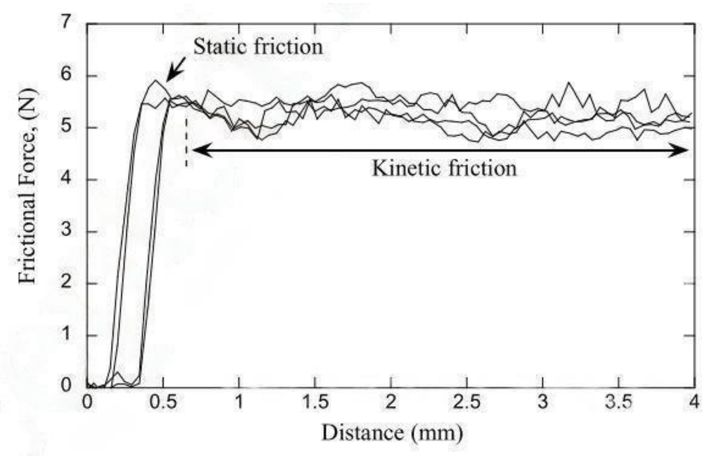


Figure 2.11: Frictional force profile along the sliding distance of 4.0 mm [36]

### 2.5.2 Bracket Slot Size

Another factor that influences the friction during sliding in dental brackets is bracket slot size. However, there is a slight discrepancy in the findings of some studies. Some studies concluded that the bracket slot size has no effect on friction and made no major difference in the amount of friction produced [50]. For example, there is not much difference in the mean friction generated between two-bracket slot sizes with occlusal-gingival heights of 0.022 inch and 0.018 inch with the friction force of 0.908 N and 0.905 N, respectively [47], [50]. However, other studies indicated that a smaller bracket slot size could result in a higher angle interface as it allows for more tipping, hence more binding incidence [51].

## 2.6 Effect of Ageing on Mechanical Behaviour of NiTi Wire

Commercially, ageing treatment has been used to improve the mechanical behaviour of NiTi shape memory alloys. The objective of ageing is to form as many precipitates as possible while limiting their size in order to increase the stiffness and harden the alloy. That being said, the force slope in force-deflection curve of NiTi wire can be reduced by lowering the wire stiffness through ageing treatment.

Ageing of NiTi material between 310 °C and 550 °C for 10 to 60 minutes can change the thermal transformation temperature and stress due to the gradual increase in the size and density of NiTi [13]. Moreover, by increasing the ageing temperature from 400 °C to 600 °C, the austenite finish ($A_f$) temperature can be reduced by 30 °C besides increasing the value in the stress-strain curve by 150 MPa. That being said, the force released in the deflection tests will be reduced if the NiTi wires are aged at temperatures of 500 °C and 600 °C, according to Silva et al [52].

In addition, the average $A_f$ temperature for superelastic NiTi wire ranges from 22 °C to 28 °C [24], which is lower than the normal oral temperature (36 °C). Thus, this wide temperature difference indicates that there is room for the $A_f$ temperature adjustment through ageing treatment. Liu et al mentioned that the ageing time should not exceed 60 minutes to ensure that maximal recoverability is maintained [53]. Extending the ageing duration at 500 °C beyond 60 minutes increased the residual strain from 0.3 to 5%.

### 2.6.1 Tensile Test

Typically, NiTi alloy is able to withstand a relatively high deformation strain of 8% to 10% before it experiences plastic deformation [13]. The wires exhibited excellent superelasticity at the testing temperature of 37 °C whereas the cyclic stress-strain curves of superelastic NiTi wires gradually changed and became unstable once the test temperature exceeded 50 °C [54].

In a study by Mohamad et al. [13], the wires were aged at temperatures of 400 °C to 550 °C by 25 °C increments for 30 minutes. Based on Figure 2.12(a), it can be seen that the stress level decreased as the ageing temperature decreased. The wires aged at 475 °C and below exhibited a shorter strain plateau, which indicated that the deformation to the strain limit of 8% involved the elastic deformation of martensite. In addition, if the loading force exceeds the elastic limit, the wire is likely to display some permanent deformation, resulting in a slight residual deflection at the end of the unloading [40]. As shown in Figure 2.12(b), the wire aged at 450 °C with 0.4% residual strain produced the best superelasticity. This is due to the coherent precipitates, which resulted in producing the least irreversible strain with 0.4%. On the contrary, due to insufficient ageing temperature, the precipitates did not obtain the necessary size for good recoverability in samples aged at 400 °C and 425 °C.

It is observed that the residual strain ranges from 0.4 to 0.95%. For ageing temperatures of 450 °C and above, the residual strain increased as the ageing temperature increased. The recoverability of the alloy therefore reduced at higher ageing temperatures [12]. This claim is supported in another research [55]. Consistent with Mohamad et al. [13], Nashrudin et al. [55] also discovered that the critical stress and residual strain decreased with increasing ageing temperature, ranging from 400 °C to 450 °C with the ageing duration of 10 minutes.

Despite that, all of the aged wires demonstrated good superelasticity with less than 1% residual strain, indicating that 400 °C to 550 °C for 30 minutes are able to generate appropriate precipitates that are coherent in reinforcing the NiTi matrix. Regardless, higher ageing temperature above 450 °C should be avoided as it decreases the recoverability of NiTi alloy and this is due to some plasticity experienced by the wires as shown in Figure 2.12(b).

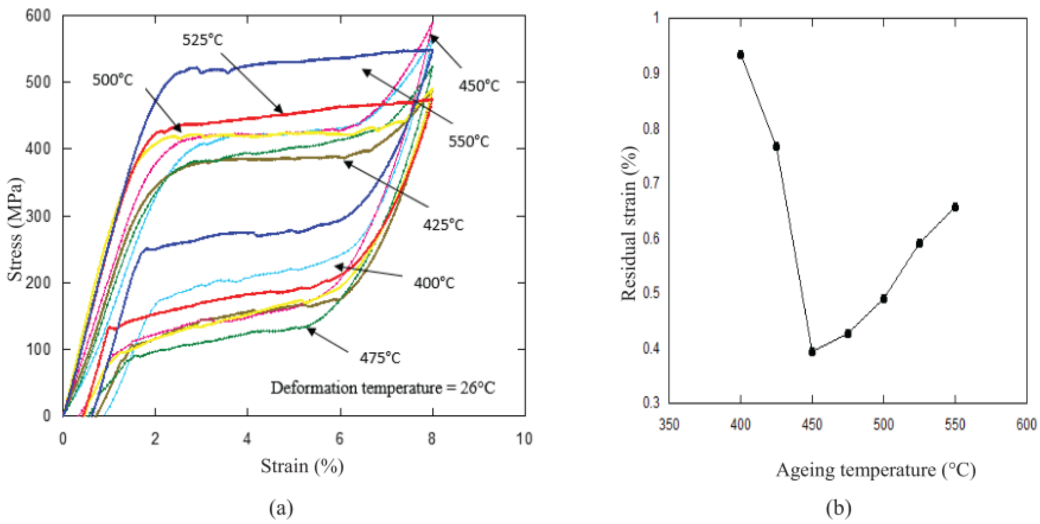


Figure 2.12: Tensile deformation behaviour of aged wires at 26 °C (a) stress-strain curves and (b) residual strain upon unloading [13]

Variations of force magnitude produced by the specimens after ageing are mainly due to the formation of precipitates. It has been found that precipitate with high coherency increases the transformation temperature and decreases the critical stress required to produce martensite phase during deformation [23]. As the temperature and time of the ageing increase, so does the size of the precipitate as illustrated in Figure 2.13.

It was found that 400 °C to 500 °C were the optimum ageing temperature to achieve complete superelasticity due to the presence of $Ni_4Ti_3$ precipitates in the matrix alloy [11], [56]. These fine and coherent precipitates strengthen the alloy and in turn, prevent slip upon loading. Omrani et al. [57] revealed that raising the temperature at the same ageing time resulted in coarser precipitation due to increased coefficient diffusion at higher temperatures. Gall et al. [56] reported that the size of $Ni_4Ti_3$ precipitates increased as the ageing temperature increased from 400 °C to 500 °C. Shu Yong et al. [12] also discovered that specimens aged at 450 °C produced fine and dense $Ni_4Ti_3$ precipitates.

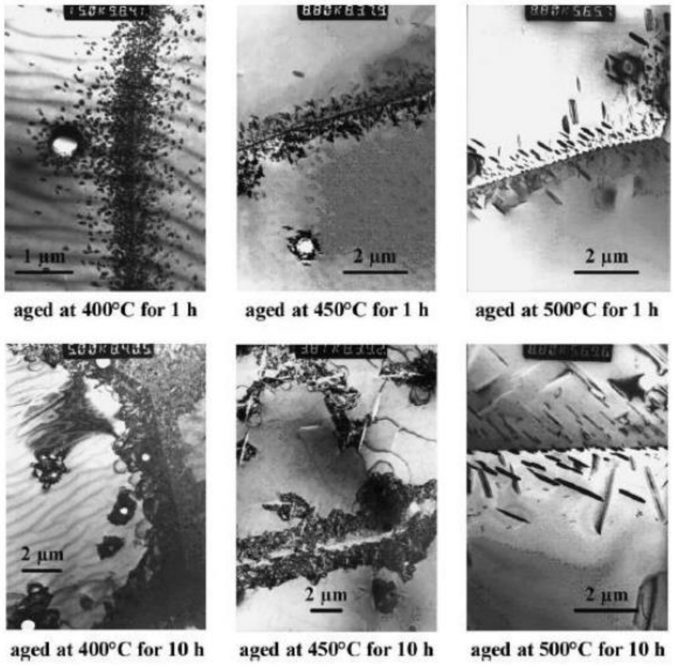


Figure 2.13: TEM micrographs of NiTi alloy after 1 hour and 10 hours of ageing at 400 °C, 450 °C, and 500 °C [58]

Other than the formation of precipitates, the variations of force magnitude exhibited by aged NiTi alloy also depend on the thermal transformation behaviour. Figure 2.14 depicts the progression of the martensitic phase transformation behaviour of wire in differential scanning calorimetry (DSC) after 30 minutes of ageing at different temperatures. The temperature at which the phase transformation between austenite (A), martensite (M), and rhombohedral (R) structure occurs is represented by the peaks. In general, the thermal transformation temperature differed and the number of transformation peaks increased when the archwires were aged at different temperatures. For example, the temperature required for full austenite transformation has climbed from 20 °C for as-received specimens to 32 °C for 490 °C aged specimens.

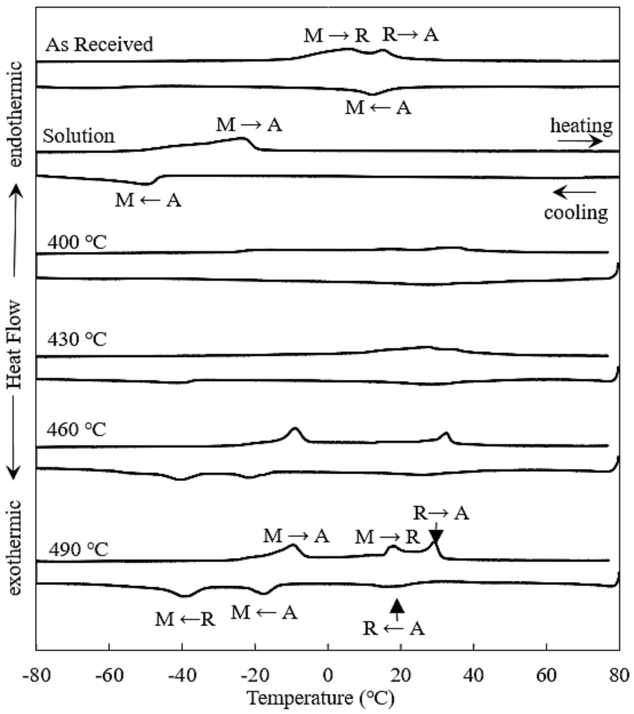


Figure 2.14: DSC curves of NiTi alloy after ageing for 30 minutes at different temperatures, ranging from 400 °C to 490 °C [59]

### 2.6.2 Three-Point Bending

There were very limited researches on the effect of ageing on NiTi alloy in three-point bending. Most of the studies mainly focused on the mechanical behaviour of aged NiTi alloy in tensile test.

Based on [60], the wires were aged at 400 and 500 °C for 30 and 60 minutes, which were then proceeded with three-point bending with the maximum deflection of 7 mm to study the superelastic behaviour of aged NiTi alloy. As seen in Figure 2.15, at this deflection, all samples showed nearly complete recovery, indicating that all samples exhibited superelastic behaviour. It is also observed that the loading and unloading plateaus grew with increasing ageing time. Ageing duration of 60 minutes produced higher force plateaus upon loading and unloading compared to 30 minutes. Both ageing temperatures showed the same force behaviour. However, the study discovered that the plateaus decreased with increasing ageing temperature as wire aged at 500 °C exhibited lower force plateaus than that of 400 °C. Furthermore, the mechanical hysteresis reduces as the ageing temperature rises. Findings of the study concluded that ageing at 500 °C for 30 and 60 minutes is the ideal ageing parameters for NiTi alloy as it has high deformation recovery and superelasticity at ambient and body temperatures, respectively.

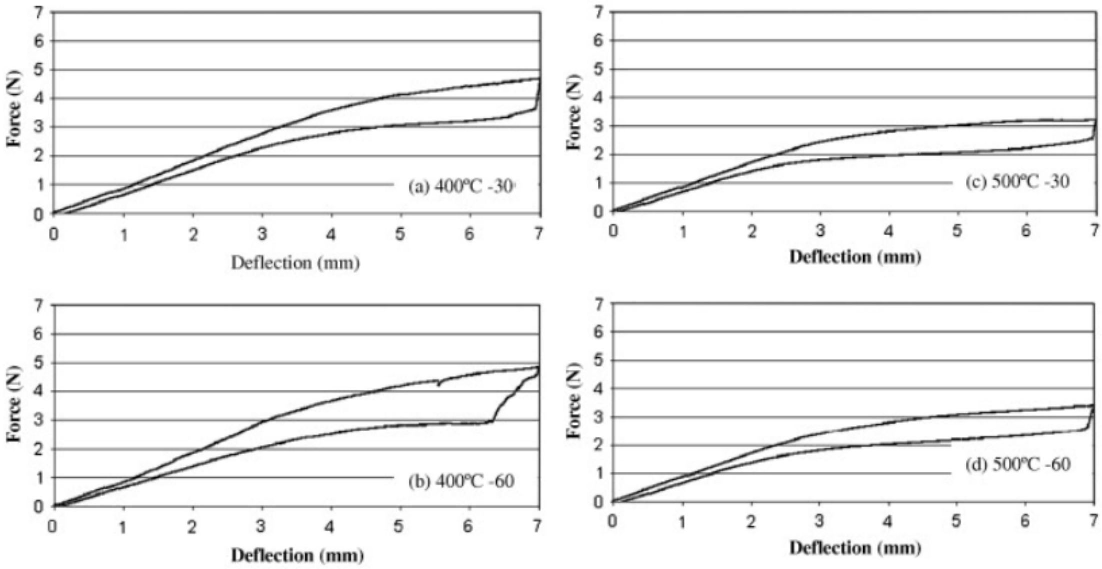


Figure 2.15: Force-deflection curves of three-point bending performed after ageing at different parameters [60]

## 2.7 Effect of Bending and Bracket Setting on Force-Deflection Behaviour

Several researchers have discovered that the bracket settings, which includes interbracket distance, bending temperature, and deflection magnitude, affect the resistance to sliding [61], [62]. In consequence, the force applied to the teeth by NiTi archwires may be affected by bracket setting, particularly the unloading force.

A study by Phermsang-ngarm et al. [63] found that different deflection magnitudes influenced the load-deflection behaviour of NiTi wires. The experiment was conducted on 0.012-inch superelastic NiTi archwire at 37 °C with interbracket span of 8 mm. It is shown in Figure 2.16 that the maximum loading forces gradually increased as deflection increased and the highest mean loading force is at 4-mm deflection with the magnitude of 6.41 N. Moreover, negative gradient slopes can be observed upon unloading at the deflection of 3 and 4 mm. Deflection at 2 mm, on the other hand, showed a similar trend except during the deactivation, which remained flat.

However, the presence of elastomeric ligature had to be taken into account as this certainly had an impact on the wire. As per mentioned in [8], the ligatures caused