Bending and Phase Transformation Properties of a Force Gradient Nickel-Titanium Orthodontic Wire

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BENDING AND PHASE TRANSFORMATION PROPERTIES OF A FORCE
GRADIENT NICKEL-TITANIUM ORTHODONTIC WIRE

by

Brinda Shah, D.D.S.

A Thesis submitted to the Faculty of the Graduate School,
Marquette University,
in Partial Fulfillment of the Requirements for
the Degree of Master of Science

Milwaukee, Wisconsin
August 2021
Objective: Conventional nickel-titanium (NiTi) archwires have widespread application in orthodontics due to their unique properties and ability to move teeth with light, continuous forces, but a shortcoming is that they apply near-uniform force along anterior and posterior teeth. Since the force needed to effectively move teeth is a function of the root surface area and bony support, these archwires cause single-rooted teeth to move more than the multi-rooted molars. With the advent of laser-engineered force gradient NiTi orthodontic archwires, individual forces may be applied to different segments or individual teeth of the arch to maximize tooth movement. The aim of this study was to evaluate force distribution and phase transformation properties of a seven-zone force gradient nickel-titanium orthodontic archwire.

Materials and Methods: SmartArch™ Laser Engineered CuNiTi archwires (0.016”; Ormco) with seven distinct force zones were investigated. They were compared to the conventional 27°C Superelastic CuNiTi archwires (0.016”; Ormco). A three-point bend test was performed to evaluate mechanical properties and a differential scanning calorimeter (DSC) was used to determine the austenite-finish (A_f) temperature for each wire segment from the central incisor to second molar. Data were analyzed via a two-way ANOVA with factors of wire and zone.

Results: The 27°C CuNiTi archwire exhibited an overall significantly different (p<0.05) A_f than the SmartArch™ wire. As expected, the CuNiTi wire exhibited statistically equivalent (p>0.05) A_f temperatures along each zone, whereas SmartArch™ showed significantly different (p<0.05) A_f temperatures along each zone, but in a pattern not as expected based upon advertised force values. There were significant differences (p<0.05) for almost all the bending parameters when comparing CuNiTi wires to SmartArch™ wires. As expected, the SmartArch™ wires had statistically different (p<0.05) force values along each zone, but in a pattern inconsistent with advertised force values. The CuNiTi wires also had statistically different (p<0.05) force values among the zones, which was also inconsistent with its advertised properties.

Conclusion: Based upon the different A_f temperatures and force values, it appears thermal treatment via laser can alter the metallurgy, phase transformation, and mechanical properties of the SmartArch™ orthodontic archwire, but this commercial wire may not exhibit the advertised force distribution.
ACKNOWLEDGEMENTS

Brinda Shah, D.D.S.

I would like to sincerely thank my thesis mentor, Dr. David Berzins, for his guidance, patience, and support throughout this research project. Dr. Berzins has been a mentor to me since my very first year of dental school, and I have the utmost respect for him as a professional. I would also like to thank the other members of my thesis committee, Dr. Dawei Liu and Dr. Ghada Nimeri, for their mentorship and assistance with my thesis. I would like to thank Marquette University and the Graduate School for their dedication to excellence.

I would like to extend my deepest gratitude and immense appreciation to my amazing parents and sweet sister for being my rocks and support system at every step of my life. Your love, compassion, and selflessness never fail to get me through. Finally, from the bottom of my heart, I would like to thank my fiancé, Pulin Joshi, for his unwavering support throughout residency and beyond. Thank you for being my guiding light always so full of hope.
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CHAPTER 1: INTRODUCTION

Over the years, clinicians have performed orthodontic treatment to correct various malocclusions through the application of fixed appliance therapy. Ideally, these appliances should transmit light, continuous forces to the dentition for optimal tooth movement as the surrounding alveolar bone and periodontium remodels (Kusy, 1997; Proffit et al., 2018). One of the main components of the fixed appliance system is the orthodontic archwire.

Orthodontic tooth movement of misaligned teeth is often performed using an archwire that is engaged to the teeth via bonded brackets (Andreasen, 1977). During comprehensive orthodontic treatment, a progression of archwires is typically used to achieve specific orthodontic goals. The beginning stage of treatment requires archwires that aid in the initial leveling and aligning of the dentition in order to correct crowding, misalignment, and rotations that may be present (Wang et al., 2018). It is crucial that this first phase of treatment is completed with quality results in order to obtain proper esthetics, function, occlusion, and overall stability for the patient (Quintão et al., 2009). The archwires used to achieve this initial orthodontic correction should ideally have the properties of low stiffness, long range of action, excellent strength, high springiness, and reasonable cost. Additionally, the ideal archwire should produce adequate biological responses by transmitting light, continuous forces and thereby reducing or eliminating adverse side effects to the periodontium during tooth movement such as pain, root resorption, hyalinization, and necrosis (Proffit et al., 2018).

Nickel-Titanium (NiTi) archwires are the most common type of archwires used in this initial stage of treatment for aligning teeth (Bellini et al., 2016). Due to this, NiTi
wires have had important use in orthodontic applications and have been evolving since their introduction in the early 1970s (Andreasen and Hilleman, 1971; Andreasen and Brady, 1972). These wires have the unique properties of superelasticity and shape memory, which make them a favorable initial archwire due to their superior working range and ability to resist permanent deformation while also maintaining a long state of activation. Additionally, NiTi wires have a low modulus of elasticity and therefore a low deflection load—this allows them to provide a light magnitude but constant duration of force to teeth without as many reactivations (Gatto et al., 2013).

The success of the present day NiTi archwire is primarily attributed to its unique properties of superelasticity and shape memory. Superelasticity describes the load-deflection curve with a horizontal region during deactivation, often called the plateau (Bellini et al., 2016). The property of superelasticity allows NiTi alloys to exist in two different crystal structures: martensite and austenite. During the austenite phase, the crystal structure is body-centered cubic; during the martensite phase, the crystal structure is monoclinic (Berzins and Roberts, 2010). In response to temperature changes or mechanical stress, the crystal structure of NiTi alloys undergoes a phase transformation. Each NiTi alloy has a unique temperature during which the crystal phase transformation occurs—this is called the temperature transition range (TTR). Generally, at lower temperatures and higher stress, the martensite phase is stable; at higher temperatures and lower stress, the austenite phase is stable (Kusy and Whitley, 2007). Another unique feature of NiTi wires is their shape memory effect. When the temperature is below the TTR, the NiTi alloy exists in the martensite phase and can plastically deform. When the temperature is above the TTR, the NiTi alloy exists in the austenite phase and can return
to its original archform shape due to the shape memory effect (Santoro et al., 2001). These unique characteristics allow NiTi wires to be effective in the initial leveling and aligning of teeth.

The amount of force needed to optimally align teeth during orthodontic treatment is a function of the root surface area and the supporting bone. Therefore, teeth with lesser root surface area, such as single-rooted incisors, should ideally have less force applied to them compared to teeth with greater root surface area, such as the multi-rooted molars. However, conventional NiTi wires deliver a near-constant force across the entire arch without much distinction between the different tooth segments (Lombardo et al., 2019). With metallurgical advances, NiTi alloys can undergo phase modifications or microstructure alterations via the addition of another component, heat treatment, or laser modification to optimize the metal alloy’s mechanical features. Heat treating and laser engineering NiTi alloys has been used to alter the transition temperature of the NiTi alloy and the force levels along a single archwire (Bellini et al., 2016). This technology has led to the emergence of variable force NiTi wires to optimize the amount of force delivered to the different regions or tooth segments of the arch and therefore improve tooth movement efficiency and overall treatment time.

Most commonly, three-force zone archwires with distinct force levels for the incisor, premolar, and molar regions have become available. The philosophy of these wires involves progressive force delivery from the anterior to posterior regions to move each area of the arch the same amount per activation. However, though these archwires are customized to deliver forces specific to the three regions of the dentition, they do not
consider the proper force needed for each individual tooth of the arch (Cherneva and Petrunov, 2020).

Recently, a laser-engineered seven-force zone archwire, SmartArch™ by Ormco, has become available in the market and is advertised to deliver ideal physiological forces to each individual tooth of the arch by accounting for average root surface area, bone support, and interbracket distance (Roberts et al., 2019). Due to the more recent development of this archwire, there has not been enough research to support the claims made by the manufacturers about the distinct force zones. Therefore, the aim of this study was to evaluate the force distribution and phase transformation properties of this seven-force gradient nickel-titanium orthodontic archwire to understand the advertised force levels.
CHAPTER 2: LITERATURE REVIEW

The dental specialty of orthodontics involves diagnosing, treatment planning, and correcting various types of malocclusions that exist in the population. Orthodontic treatment goals include obtaining a functional occlusion, adequate esthetics, stability, and overall facial balance. The appliances and methods used to correct malocclusions and achieve these goals date back to 1000 B.C. while more contemporary styles date back to the late 1800s when the Father of Modern Orthodontics, Edward Angle, was a prominent figure in the field (Proffit et al., 2018). One of the appliances that has been extensively used for orthodontic treatment is the edgewise appliance. This appliance has evolved significantly over the years to become the modern-day fixed appliance that is used currently.

These fixed appliances should ideally transmit light, continuous forces to the dentition in order to produce optimal tooth movement and eliminate adverse side effects including pain and damage to the periodontium. One of the main components of this appliance is the orthodontic archwire which transmits forces to the teeth through stored energy when engaged to bonded brackets (Eliades et al., 2000). Throughout the years, orthodontic archwires have gone through multiple evolutions to achieve more ideal properties. Orthodontic treatment typically begins with more flexible archwires during the beginning phases of treatment—the leveling and aligning of teeth—while stiffer archwires are used in the later stages of treatment—the detailing and finishing of teeth (Quintão and Brunharo, 2009).

Nickel-Titanium archwires have widespread use in orthodontics and are primarily used during this initial phase of leveling and aligning the dentition. This is due to their
low, constant force delivery and long elastic working range, which makes them a favorable archwire for orthodontic treatment (Gatto et al., 2013).

The History of Nickel-Titanium and Its Properties

Nickel-Titanium was first developed in the early 1960s at the U.S. Naval Ordnance Laboratory by William Beuhler and his colleagues for the space program (Brantley, 2020). Though this metal alloy was not initially created for orthodontic purposes, its unique properties, shape memory capacity, and ability to deliver light, continuous forces quickly became of interest. After a few years, George Andreasen introduced NiTi wires to the field of orthodontics through his innovative studies in the early 1970s (Andreasen & Hilleman, 1971). Since its early development, the NiTi archwire has quickly progressed to becoming the wire of choice during the initial leveling and aligning phase of orthodontic treatment (Gatto et al., 2013).

The first generation of NiTi archwires was coined “Nitinol” wires and was based on its composition (Nickel and Titanium) and its origin (Naval Ordnance Laboratory). These original non-superalloy Nitinol™ wires were essentially equiatomic nickel and titanium alloys and were first available through the Unitek Corporation (Brantley, 2020). This earlier version of NiTi archwires was characterized by a low modulus of elasticity and long range of action compared to the stainless steel archwires. These properties are largely due to the phase transformation ability of NiTi alloys—they can reversibly change between martensite, which is stabilized at low temperatures and high stress, and austenite, which is stabilized at high temperatures and low stress. However, compared to its modern-day NiTi counterpart, these first generation NiTi wires did not exhibit the true
shape memory effect due to cold working during manufacturing and were therefore only stabilized in the martensitic crystal structure (Kusy, 1997).

The next big evolution in NiTi orthodontic products involved superelastic NiTi wires which were successful due to their high springback, continuous force delivery, and low deflection rate. Fujio Miura and colleagues developed the superelastic Japanese NiTi which was marketed as Sentalloy® by GAC while Charles Burstone and colleagues introduced the superelastic Chinese NiTi which was marketed as Ni-Ti™ byOrmco (Brantley, 2020). These new NiTi wires were austenitic-active and could switch between the austenite and martensite phases depending on the phase transformation within the alloy. These superelastic wires are austenite at room temperature, return to martensite when the wire has applied stress, and then switch back to austenite once the applied load is removed (Bellini et al., 2016).

After additional technological advancements, the first shape memory NiTi archwires entered the market. These third generation NiTi wires, marketed as Neo Sentalloy® by GAC, are martensitic-active—they have a transition temperature range near body temperature and therefore exhibit the shape memory effect when placed intraorally (Brantley, 2020). Due to this, they are in their martensite phase below the transition temperature and can be easily deformed especially in cases with malaligned teeth. At the transition temperature, which would be achieved intraorally, these wires change to their austenite structure and go back to their original archform (Bellini et al., 2016).

Another third generation NiTi wire entered the market in the mid-1990s. These NiTi wires have the addition of copper (CuNiTi) to the metal alloy to enhance their
thermal and mechanical properties. These archwires have superior thermal properties to the previous NiTi wires because they have specific built-in transition temperatures at which the shape memory effect takes place. These wires were first marketed by Ormco and have three specific transition temperatures (27°C, 35°C, and 40°C), which allows orthodontists to choose the appropriate archwire for the desired treatment results (Quintão and Brunharo, 2009).

**The Advent of Multi-force Nickel-Titanium Archwires**

Historically, NiTi orthodontic archwires have produced light forces that are evenly distributed among anterior and posterior teeth. However, in most cases, these archwires cause the incisors to move more efficiently in the initial stages than the premolars and molars which cause the posterior segments to lag behind their anterior counterparts. As a result, there is often less efficient overall tooth movement which increases treatment time due to the need for more activations and archwire changes. Consequently, the demand for a more predictable and ideal force delivery system has led to the discovery of an archwire that can provide the ideal physiological force to every segment or tooth in the arch at the same time in order to produce more efficient orthodontic tooth movement and reduce overall treatment time (Olsen, 2020).

With the advent of thermally treated and laser-engineered force gradient nickel-titanium (NiTi) orthodontic archwires, individual forces can be applied to different segments or individual teeth of the arch to maximize tooth movement. This enables these wires to have “multiple memories” throughout a single archwire rather than the single functionality of conventional NiTi archwires (Khan et al., 2013).
Over the past few years, three-force zone archwires with variable transition temperatures have entered the market. These multiple shape memory alloys have been manufactured to serve as thermally active force gradient wires that apply segmental forces to the incisor, premolar, and molar regions of the dentition. Though these archwires provide light forces in the anterior region and heavier forces in the posterior region instead of the traditional uniform force distribution, they do not customize their force delivery to each tooth in the arch (Cherneva and Petrunov, 2020). Since the manufacturers did not take the physiology of each tooth root and respective periodontal ligament (PDL) into account when fabricating these three-force zone archwires, they were unable to fabricate an archwire that precisely delivers ideal physiological forces to each tooth in the arch.

Due to these shortcomings, manufacturers are using new archwire technology and materials science to develop multi-force orthodontic archwires that produce physiological force levels for each individual tooth rather than segments of the arch. These new laser-engineered force gradient NiTi archwires, marketed as SmartArch™ by Ormco, have seven distinct force zones based on the average inter-bracket distances and the physiology of the root and periodontal surfaces of every tooth in the arch (Panton et al., 2017; Olsen, 2020). Due to the recent development of this technology, further testing should be done to evaluate the force distribution of this new archwire to see if it in fact has addressed previous shortcomings and potentially could yield more efficient tooth movement, reduce treatment time, and enhance the quality of orthodontic treatment.
The Orthodontic Applications of Nickel-Titanium

The nickel-titanium alloy has found widespread application in orthodontics due to its exceptional mechanical characteristics. Specifically, the NiTi alloy is used most often as an archwire in the straight-wire fixed appliance during orthodontic treatment due to the many benefits that it offers. NiTi archwires offer biocompatibility, high strength, low stiffness, long range of action, low load-deflection ratios, superelasticity, and shape memory, which are characteristics that are advantageous during orthodontic mechanotherapy. These properties allow NiTi wires to exert forces that are light and continuous which allow them to be effective in the initial leveling and aligning phase of orthodontic treatment. The use of archwires with these properties has led to increasing patient comfort, greater time between orthodontic visits, and more treatment efficiency (Sifakakis, 2017; Aghili et al., 2015).

Additionally, these versatile NiTi wires can be altered by composition modification, heat treatment, or laser engineering during the manufacturing process. These modifications enable these wires to exhibit different transition temperatures and make it possible to have a force gradient along a single NiTi archwire. With laser engineering and heat treatment, these NiTi wires can exhibit different force levels along the archwire without changing the diameter of the wire. The result is more accurate and physiological force delivery from a single archwire rather than a progression of them (Olsen 2020).

Differential Scanning Calorimetry

In the 1960s, differential scanning calorimetry (DSC) was developed to assess the thermal properties of various material samples including changes in heat flow and phase
transition temperatures (Höhne et al., 2003). In orthodontics, DSC can be used to examine the metallurgical and phase transformation properties of nickel-titanium archwires. Specifically, DSC can be used to determine the transition temperature range of the NiTi metal alloy. While DSC primarily detects the phase transitions and changes in heat flow of the NiTi wires, this thermal analysis can also serve as an indicator of the wire’s mechanical properties (Bradley et al., 1996).

The phase transformation between austenite and martensite is largely responsible for the superelastic and shape memory capacity of NiTi wires. The three generations of NiTi wires—nonsuperelastic, superelastic, and shape memory wires—have had differing mechanical characteristics due to varying phase transformation properties. Through DSC thermal analysis, a measurement of value is the austenite-finish ($A_f$) temperature. The $A_f$ temperature is often an important indicator of the mechanical properties, bending characteristics, and force profile of the sample. In general, greater $A_f$ temperatures correspond to lower force levels (Chybowski, 2014).

**Three-Point Bend Test**

The three-point bend test is an analysis technique that was developed to assess the mechanical properties of materials. In orthodontics, this method can be used to understand the nature of orthodontic wires and their load-deflection properties, which are often the most critical variables in determining the physiological behavior of tooth movement (Sifakakis, 2017). Additionally, the three-point bend test most accurately represents the way in which the wires deform and their clinical nature compared to other mechanical tests like the tension test (Asgharnia and Brantley, 1986). More specifically, three-point bend tests can measure the activation (loading) and deactivation (unloading)
forces, the stiffness, elastic modulus, as well as elastic recovery present within the wire sample.

Therefore, since the DSC and three-point bend test can accurately assess the behavior of orthodontic wires, the objective of this study was to evaluate the mechanical and thermal properties of a single seven-force gradient nickel-titanium orthodontic archwire using these methods to understand the advertised force levels.
CHAPTER 3: MATERIALS AND METHODS

In this study, the maxillary SmartArch™ Laser Engineered CuNiTi archwires (0.016”; Ormco) with seven distinct force zones were investigated to evaluate their thermomechanical properties. They were compared to the conventional maxillary 27°C Superelastic Copper NiTi archwires (0.016”; Ormco) without any distinct force zones which served as a non-variable force control group. For both wires, seven distinct wire zones were evaluated—the central incisor, lateral incisor, canine, 1st premolar, 2nd premolar, 1st molar, and 2nd molar regions of the arch.

To assess the mechanical and load-deflection properties of the wires, a three-point bend test was performed. The thermal properties and austenite-finish temperature ($A_f$) was determined by differential scanning calorimetry (DSC). Both methods were performed for each wire segment from the central incisor to second molar regions. The segments were sectioned so that the middle of each segment was coincident with the middle of each zone.

**Differential Scanning Calorimetry**

The SmartArch™ and CuNiTi wires that were used in this study were in the condition they were received from the manufacturer. Each wire was sectioned into seven segments representing each of the seven tooth zones: central incisor, lateral incisor, canine, 1st premolar, 2nd premolar, 1st molar, and 2nd molar (n=7/wire/zone).

Segments (4 mm) of each zone were obtained by sectioning areas along the SmartArch™ archform seen in Figure 1 and along the CuNiTi Tru-Archform as seen in Figure 2. The middle of each segment was coincident with the middle of each tooth zone. Each 4 mm segment was precisely sectioned from the archwire using a low-speed, wet...
diamond saw (Isomet, Buehler Ltd., Lake Bluff, Illinois) in order to prevent inadvertent heating and structural damage from mechanical stresses. This Isomet diamond saw is pictured in Figure 3. The same investigator cut all wire segments.

The 4 mm wire samples were weighed to the nearest 0.01 mg using an electronic scale (Mettler-Toledo Inc., Columbus, Ohio) as seen in Figure 4. The samples were then placed and sealed into an aluminum crucible. An empty aluminum crucible was also placed in the DSC apparatus to serve as a reference. Each wire sample was then thermally scanned using a Differential Scanning Calorimeter (Model 822e, Mettler-Toledo Inc., Columbus, Ohio) as pictured in Figure 5.

Using the DSC, the wire samples were heated from -100°C to 100°C using the built-in furnace and then cooled from 100°C to -100°C using liquid nitrogen at 10°C/minute. This technique resulted in heating and cooling curves (thermograms) specific to each wire type and tooth zone. These thermograms were analyzed using specific software from the DSC manufacturer. The heating onset temperatures, heating endset temperatures, heating enthalpy, cooling enthalpy, cooling onset temperatures, and cooling endset temperatures were obtained from the thermograms.
Figure 1: The SmartArch™ Natural archform used to obtain 4 mm wire segments for the DSC technique (not pictured to scale).

Figure 2: The 27°C Superelastic CuNiTi Tru-Archform used to obtain 4 mm wire segments for the DSC technique (not pictured to scale).
Figure 3: The Isomet™ low-speed, water-coolant diamond saw used to section archwires for DSC testing.

Figure 4: Mettler-Toledo analytical balance used to weigh archwire segments for DSC testing.
Three-Point Bend Test

The three-point bend test was performed to assess the mechanical properties, especially the load-deflection characteristics, of the SmartArch™ and CuNiTi wires. This test was used to analyze the bending forces for a particular amount of deflection for the two wires. It was performed using the Instron universal testing machine (Model 5500R, Instron Corp., Canton, MA) as pictured in Figure 6. This was set up in accordance with the ADA Specification No. 32 with the exception of the lower fixture span being 14 mm instead of the 10 mm standard due to fixture limitations. The surrounding temperature was maintained at 37±2°C using a portable heater as seen in Figure 6.

The SmartArch™ and CuNiTi wires that were used in this study were in the condition they were received from the manufacturer. Twenty wire segments of each tooth
zone (central incisor, lateral incisor, canine, 1st premolar, 2nd premolar, 1st molar, and 2nd molar) of each wire type (SmartArch™ and CuNiTi) were tested (n=20/wire/zone). Each segment was 20 mm in length. These 20 mm segments of each zone were obtained by sectioning areas along the SmartArch™ archform seen in Figure 1 and along the CuNiTi Tru-Archform seen in Figure 2. The middle of each segment was coincident with the middle of each tooth zone. Each 20 mm segment was carefully sectioned from the archwire using a wire cutter. The same investigator cut all wire segments.

The 20 mm segments were positioned flat onto the three-point bend test fixture as seen in Figure 7 and were deflected at a rate of 2 mm/minute to a deflection of 3.1 mm as pictured in Figure 8. The segments were subsequently unloaded at the same rate of 2 mm/minute to their original flat position. The loading (activation) and unloading (deactivation) forces were recorded directly onto a computer software (Merlin, Instron Corp.). The linear slope of the bending curves was obtained from the measured data and represented the stiffness of the wire material. The stiffness was then used to calculate the bending modulus. Additionally, the activation and deactivation force at 1 mm, 2 mm, and 3 mm of deflection was obtained from the data.
Figure 6: Instron Universal Testing Machine Model 5500R used for the three-point bend test with the portable heater in the background.

Figure 7: Wire segment positioned flat on the Instron three-point bend test fixture.
Statistical Analysis

The three-point bend test and differential scanning calorimetry data were analyzed with a two-way ANOVA with factors of wire (SmartArch™ or CuNiTi) and zone (central incisor, lateral incisor, canine, 1st premolar, 2nd premolar, 1st molar, 2nd molar). A post-hoc Tukey test was performed to analyze any differences between the seven zones for both the SmartArch™ and CuNiTi wires. The significance level was set at p<0.05.

Statistical analysis was performed using SPSS software (SPSS Inc., Chicago, Illinois).

Figure 8: Wire segment loading during the three-point bend test.
CHAPTER 4: RESULTS

Differential Scanning Calorimetry

DSC thermal parameter data (heating enthalpy, heating onset temperature, heating endset temperature, cooling enthalpy, cooling onset temperature, cooling endset temperature) were analyzed via two-way ANOVA with factors of wire (SmartArch™ or CuNiTi) and zone (central incisor, lateral incisor, canine, 1<sup>st</sup> premolar, 2<sup>nd</sup> premolar, 1<sup>st</sup> molar, 2<sup>nd</sup> molar). There were significant differences (p<0.05) with respect to all DSC thermal parameters between the wires and among the zones, with the exception that cooling enthalpy was not significantly (p>0.05) different between wires. A significant interaction (p<0.05) was also observed between wire and zone. Due to this interaction, the DSC thermal parameter data were analyzed with respect to zone within each wire using one-way ANOVA followed up with a post-hoc Tukey test, if indicated.

For CuNiTi, one-way ANOVA showed no significant differences (p>0.05) between zones with respect to heating enthalpy, heating onset temperature, heating endset temperature, and cooling onset temperature. Significant differences (p<0.05) between zones with respect to cooling enthalpy and cooling endset temperature were observed, however, and the individual groupings are displayed in Table 1 via different letters.

For SmartArch™, one-way ANOVA showed significant differences (p<0.05) between zones with respect to all heating and cooling parameters with individual zone differences shown in Table 2 via different letters.
Table 1. DSC measured temperature and enthalpy changes for phase transformations during heating and cooling of 0.016” 27°C CuNiTi wires.

<table>
<thead>
<tr>
<th>Zone</th>
<th>Onset Temperature (°C)</th>
<th>Endset Temperature (°C)</th>
<th>Enthalpy (J/g)</th>
<th>Onset Temperature (°C)</th>
<th>Endset Temperature (°C)</th>
<th>Enthalpy (J/g)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Central</td>
<td>5.4±0.24</td>
<td>24.1±0.39</td>
<td>-12.8±0.95</td>
<td>8.8±0.34</td>
<td>-18.5±0.64</td>
<td>11.0±0.59 ABC</td>
</tr>
<tr>
<td>Lateral</td>
<td>5.6±0.42</td>
<td>24.0±0.36</td>
<td>-13.2±0.6</td>
<td>8.6±0.27</td>
<td>-19.2±0.64</td>
<td>11.9±0.88 A</td>
</tr>
<tr>
<td>Canine</td>
<td>5.5±0.46</td>
<td>24.2±0.53</td>
<td>-13.2±0.8</td>
<td>8.2±0.41</td>
<td>-17.7±1.62</td>
<td>9.6±0.84 C</td>
</tr>
<tr>
<td>1st Premolar</td>
<td>5.4±0.36</td>
<td>23.9±0.50</td>
<td>-13.0±0.52</td>
<td>8.2±0.60</td>
<td>-17.8±0.56</td>
<td>10.4±1.22 BC</td>
</tr>
<tr>
<td>2nd Premolar</td>
<td>6.5±3.18</td>
<td>23.6±0.78</td>
<td>-13.1±0.36</td>
<td>8.2±0.33</td>
<td>-17.6±.72</td>
<td>9.9±1.26 BC</td>
</tr>
<tr>
<td>1st Molar</td>
<td>5.5±0.34</td>
<td>24.0±0.50</td>
<td>-13.0±0.81</td>
<td>8.5±0.26</td>
<td>-19.8±1.41</td>
<td>12.1±0.79 A</td>
</tr>
<tr>
<td>2nd Molar</td>
<td>5.4±0.43</td>
<td>24.0±0.52</td>
<td>-12.9±0.67</td>
<td>8.5±0.40</td>
<td>-18.9±0.84</td>
<td>11.2±0.56 AB</td>
</tr>
</tbody>
</table>

Different letters indicate a statistically significant (p<0.05) difference exists between wires for a given measure. If no letter is given for a column, no significant (p>0.05) differences were found.

Table 2. DSC measured temperature and enthalpy changes for phase transformations during heating and cooling of 0.016” SmartArch™ wires.

<table>
<thead>
<tr>
<th>Zone</th>
<th>Onset Temperature (°C)</th>
<th>Endset Temperature (°C)</th>
<th>Enthalpy (J/g)</th>
<th>Onset Temperature (°C)</th>
<th>Endset Temperature (°C)</th>
<th>Enthalpy (J/g)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Central</td>
<td>-2.1±0.93</td>
<td>19.5±0.53</td>
<td>-10.8±1.31</td>
<td>7.0±0.45</td>
<td>-28.6±1.75</td>
<td>9.2±1.03 CD</td>
</tr>
<tr>
<td>Lateral</td>
<td>9.7±2.85</td>
<td>22.1±2.80</td>
<td>-15.0±0.46</td>
<td>7.8±1.07</td>
<td>-9.7±5.53</td>
<td>16.5±0.61 A</td>
</tr>
<tr>
<td>Canine</td>
<td>4.1±1.48</td>
<td>21.4±1.37</td>
<td>-12.8±0.67</td>
<td>8.8±2.24</td>
<td>-20.7±2.28</td>
<td>11.2±1.49 BC</td>
</tr>
<tr>
<td>1st Premolar</td>
<td>3.9±2.05</td>
<td>22.7±2.65</td>
<td>-12.9±0.71</td>
<td>8.4±1.71</td>
<td>-16.9±3.99</td>
<td>13.0±1.16 B</td>
</tr>
<tr>
<td>2nd Premolar</td>
<td>0.4±2.82</td>
<td>21.4±0.81</td>
<td>-11.5±0.76</td>
<td>7.7±0.69</td>
<td>-21.5±4.78</td>
<td>9.8±1.25 CD</td>
</tr>
<tr>
<td>1st Molar</td>
<td>-2.4±0.91</td>
<td>18.3±0.38</td>
<td>-10.8±0.76</td>
<td>4.9±0.43</td>
<td>-29.6±2.54</td>
<td>7.9±1.11 D</td>
</tr>
<tr>
<td>2nd Molar</td>
<td>-2.2±0.96</td>
<td>18.3±0.23</td>
<td>-11.0±0.48</td>
<td>4.9±0.66</td>
<td>-28.6±1.53</td>
<td>8.2±1.40 D</td>
</tr>
</tbody>
</table>

Different letters indicate a statistically significant (p<0.05) difference exists between wires for a given measure. If no letter is given for a column, no significant (p>0.05) differences were found.
Figures 9-12 represent the comparison of the heating and cooling thermograms of each tooth zone of each wire type (CuNiTi and SmartArch™). Figures 13-26 display the representative thermogram of each individual tooth zone of each wire type (CuNiTi and SmartArch™).

**Figure 9:** Comparison of heating curves of all zones of CuNiTi.
Figure 10: Comparison of cooling curves of all zones of CuNiTi.

Figure 11: Comparison of heating curves of all zones of SmartArch™.
Figure 12: Comparison of cooling curves of all zones of SmartArch™.

Figure 13: Representative thermogram of the central incisor zone of CuNiTi.
**Figure 14:** Representative thermogram of the lateral incisor zone of CuNiTi.

**Figure 15:** Representative thermogram of the canine zone of CuNiTi.
Figure 16: Representative thermogram of the 1st premolar zone of CuNiTi.

Figure 17: Representative thermogram of the 2nd premolar zone of CuNiTi.
Figure 18: Representative thermogram of the 1st molar zone of CuNiTi.

Figure 19: Representative thermogram of the 2nd molar zone of CuNiTi.
Figure 20: Representative thermogram of the central incisor zone of SmartArch™.

Figure 21: Representative thermogram of the lateral incisor zone of SmartArch™.
Figure 22: Representative thermogram of the canine zone of SmartArch™.

Figure 23: Representative thermogram of the 1st premolar zone of SmartArch™.
Figure 24: Representative thermogram of the 2nd premolar zone of SmartArch™.

Figure 25: Representative thermogram of the 1st molar zone of SmartArch™.
Figure 26: Representative thermogram of the 2nd molar zone of SmartArch™.

Three-Point Bend Test

Bending data (percent recovery, activation stiffness, activation modulus, activation force at 1 mm, activation force at 2 mm, activation force at 3 mm, deactivation stiffness, deactivation modulus, deactivation force at 1 mm, deactivation force at 2 mm, deactivation force at 3 mm) were analyzed via two-way ANOVA with wire (SmartArch™ or CuNiTi) and zone (central incisor, lateral incisor, canine, 1st premolar, 2nd premolar, 1st molar, 2nd molar) as factors. There were significant differences (p<0.05) with respect to 17 of the 22 bending parameters between the wires and among the zones, with the exceptions being percent recovery for wire and zone, and deactivation stiffness/modulus/forces at 2 and 3 mm. A significant interaction (p<0.05) was also observed between wire and zone for all parameters. Due to this interaction, the bending...
parameter data were analyzed with respect to zone within each wire using one-way ANOVA followed up with a post-hoc Tukey test, if indicated.

For CuNiTi, one-way ANOVA showed significant differences (p<0.05) between zones with respect to all bending parameters with the individual groupings displayed in Tables 3 and 4 via different letters.

For SmartArch™, one-way ANOVA showed no significant differences (p>0.05) between zones with respect to percent recovery, whereas all other bending parameters were significantly different (p<0.05) between zones. Individual groupings for SmartArch™ are similarly displayed in Tables 5 and 6 via different letters.

Table 3. Bending values during activation of 0.016” 27°C CuNiTi wires.

<table>
<thead>
<tr>
<th>Wire</th>
<th>Stiffness (g/mm)</th>
<th>Modulus (GPa)</th>
<th>Force at 1 mm (g)</th>
<th>Force at 2 mm (g)</th>
<th>Force at 3 mm (g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Incisor</td>
<td>216±7 BC</td>
<td>118±4 BC</td>
<td>131±9 B</td>
<td>138±10 D</td>
<td>136±9 C</td>
</tr>
<tr>
<td>Lateral Incisor</td>
<td>221±5 AB</td>
<td>121±3 AB</td>
<td>131±9 B</td>
<td>140±8 D</td>
<td>141±1 C</td>
</tr>
<tr>
<td>Canine</td>
<td>225±6 A</td>
<td>123±3 A</td>
<td>143±7 A</td>
<td>151±8 B</td>
<td>144±8 BC</td>
</tr>
<tr>
<td>1st Premolar</td>
<td>219±4 ABC</td>
<td>120±2 ABC</td>
<td>142±3 A</td>
<td>148±2 BC</td>
<td>141±3 C</td>
</tr>
<tr>
<td>2nd Premolar</td>
<td>214±4 C</td>
<td>117±2 C</td>
<td>141±4 A</td>
<td>152±5 B</td>
<td>152±6 B</td>
</tr>
<tr>
<td>1st Molar</td>
<td>218±6 BC</td>
<td>119±3 BC</td>
<td>146±8 A</td>
<td>161±9 A</td>
<td>164±8 A</td>
</tr>
<tr>
<td>2nd Molar</td>
<td>190±11 D</td>
<td>104±6 D</td>
<td>126±9 B</td>
<td>144±5 CD</td>
<td>152±5 B</td>
</tr>
</tbody>
</table>

Different letters indicate a statistically significant (p<0.05) difference exists between wires for a given measure.
### Table 4. Bending values during deactivation of 0.016” 27°C CuNiTi wires.

<table>
<thead>
<tr>
<th>Wire</th>
<th>DEACTIVATION</th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Stiffness (g/mm)</td>
<td>Modulus (GPa)</td>
<td>Force at 1 mm (g)</td>
<td>Force at 2 mm (g)</td>
</tr>
<tr>
<td>Central Incisor</td>
<td>196±21</td>
<td>107±12</td>
<td>80±10</td>
<td>82±9</td>
<td>115±9</td>
</tr>
<tr>
<td></td>
<td>AB</td>
<td>AB</td>
<td>B</td>
<td>D</td>
<td>D</td>
</tr>
<tr>
<td>Lateral Incisor</td>
<td>195±9</td>
<td>107±5</td>
<td>80±7</td>
<td>82±7</td>
<td>118±5</td>
</tr>
<tr>
<td></td>
<td>AB</td>
<td>AB</td>
<td>B</td>
<td>D</td>
<td>D</td>
</tr>
<tr>
<td>Canine</td>
<td>199±7</td>
<td>109±4</td>
<td>90±6</td>
<td>91±6</td>
<td>124±9</td>
</tr>
<tr>
<td></td>
<td>A</td>
<td>A</td>
<td>A</td>
<td>C</td>
<td>BCD</td>
</tr>
<tr>
<td>1st Premolar</td>
<td>193±7</td>
<td>105±4</td>
<td>89±3</td>
<td>91±2</td>
<td>121±4</td>
</tr>
<tr>
<td></td>
<td>AB</td>
<td>AB</td>
<td>A</td>
<td>C</td>
<td>CD</td>
</tr>
<tr>
<td>2nd Premolar</td>
<td>196±5</td>
<td>107±3</td>
<td>91±5</td>
<td>98±6</td>
<td>131±7</td>
</tr>
<tr>
<td></td>
<td>AB</td>
<td>AB</td>
<td>A</td>
<td>B</td>
<td>B</td>
</tr>
<tr>
<td>1st Molar</td>
<td>199±5</td>
<td>109±3</td>
<td>94±7</td>
<td>105±7</td>
<td>141±9</td>
</tr>
<tr>
<td></td>
<td>A</td>
<td>A</td>
<td>A</td>
<td>A</td>
<td>A</td>
</tr>
<tr>
<td>2nd Molar</td>
<td>186±9</td>
<td>102±5</td>
<td>82±4</td>
<td>92±4</td>
<td>130±4</td>
</tr>
<tr>
<td></td>
<td>B</td>
<td>B</td>
<td>B</td>
<td>C</td>
<td>BC</td>
</tr>
</tbody>
</table>

Different letters indicate a statistically significant (p<0.05) difference exists between wires for a given measure.
Table 5. Bending values during activation of 0.016” SmartArch™ wires.

<table>
<thead>
<tr>
<th>Wire</th>
<th>Stiffness (g/mm)</th>
<th>Modulus (GPa)</th>
<th>Force at 1 mm (g)</th>
<th>Force at 2 mm (g)</th>
<th>Force at 3 mm (g)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Incisor</td>
<td>223±4 B</td>
<td>122±2 B</td>
<td>156±3 B</td>
<td>162±3 C</td>
<td>155±3 C</td>
</tr>
<tr>
<td>Lateral Incisor</td>
<td>235±7 A</td>
<td>129±4 A</td>
<td>119±5 E</td>
<td>121±6 F</td>
<td>115±6 G</td>
</tr>
<tr>
<td>Canine</td>
<td>218±7 BC</td>
<td>119±4 BC</td>
<td>133±9 D</td>
<td>138±9 E</td>
<td>130±8 F</td>
</tr>
<tr>
<td>1st Premolar</td>
<td>231±4 A</td>
<td>127±2 A</td>
<td>136±8 CD</td>
<td>142±7 E</td>
<td>138±6 E</td>
</tr>
<tr>
<td>2nd Premolar</td>
<td>221±7 B</td>
<td>121±4 B</td>
<td>142±13 C</td>
<td>151±13 D</td>
<td>148±9 D</td>
</tr>
<tr>
<td>1st Molar</td>
<td>196±5 D</td>
<td>107±2 D</td>
<td>160±4 AB</td>
<td>190±5 A</td>
<td>197±7 A</td>
</tr>
<tr>
<td>2nd Molar</td>
<td>213±4 C</td>
<td>117±2 C</td>
<td>163±3 A</td>
<td>179±2 B</td>
<td>183±3 B</td>
</tr>
</tbody>
</table>

Different letters indicate a statistically significant (p<0.05) difference exists between wires for a given measure.
Table 6. Bending values during deactivation of 0.016” SmartArch™ wires.

<table>
<thead>
<tr>
<th>Wire</th>
<th>DEACTIVATION</th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Stiffness (g/mm)</td>
<td>Modulus (GPa)</td>
<td>Force at 1 mm (g)</td>
<td>Force at 2 mm (g)</td>
<td>Force at 3 mm (g)</td>
</tr>
<tr>
<td>Central Incisor</td>
<td>196±4</td>
<td>107±2</td>
<td>101±3</td>
<td>107±3</td>
<td>136±3</td>
</tr>
<tr>
<td>Lateral Incisor</td>
<td>200±17</td>
<td>110±9</td>
<td>50±5</td>
<td>52±5</td>
<td>93±5</td>
</tr>
<tr>
<td>Canine</td>
<td>193±11</td>
<td>106±6</td>
<td>73±11</td>
<td>76±10</td>
<td>110±9</td>
</tr>
<tr>
<td>1st Premolar</td>
<td>210±9</td>
<td>115±5</td>
<td>68±6</td>
<td>77±7</td>
<td>113±6</td>
</tr>
<tr>
<td>2nd Premolar</td>
<td>192±17</td>
<td>105±9</td>
<td>71±17</td>
<td>84±15</td>
<td>126±9</td>
</tr>
<tr>
<td>1st Molar</td>
<td>175±6</td>
<td>96±3</td>
<td>121±3</td>
<td>141±5</td>
<td>174±7</td>
</tr>
<tr>
<td>2nd Molar</td>
<td>196±5</td>
<td>107±3</td>
<td>113±2</td>
<td>126±3</td>
<td>163±3</td>
</tr>
</tbody>
</table>

Different letters indicate a statistically significant (p<0.05) difference exists between wires for a given measure.
Table 7. Percent Recovery of 0.016” 27°C CuNiTi and SmartArch™ wires.

<table>
<thead>
<tr>
<th>Wire</th>
<th>27°C CuNiTi Percent Recovery (%)</th>
<th>SmartArch™ Percent Recovery (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Central Incisor</td>
<td>99.2±1.0 A</td>
<td>98.8±0.8 A</td>
</tr>
<tr>
<td>Lateral Incisor</td>
<td>97.9±2.4 AB</td>
<td>98.8±1.0 A</td>
</tr>
<tr>
<td>Canine</td>
<td>98.8±1.0 A</td>
<td>98.3±1.0 A</td>
</tr>
<tr>
<td>1st Premolar</td>
<td>99.0±0.6 A</td>
<td>98.4±1.4 A</td>
</tr>
<tr>
<td>2nd Premolar</td>
<td>98.8±0.9 A</td>
<td>98.8±1.6 A</td>
</tr>
<tr>
<td>1st Molar</td>
<td>99.3±0.5 A</td>
<td>98.2±1.2 A</td>
</tr>
<tr>
<td>2nd Molar</td>
<td>96.8±3.5 B</td>
<td>98.8±1.1 A</td>
</tr>
</tbody>
</table>

Different letters indicate a statistically significant (p<0.05) difference exists between wires for a given measure.

Figures 27-28 represent the comparison of the bending curves of each tooth zone of each wire type (CuNiTi or SmartArch™). Figures 13-26 display the representative bending curve of each individual tooth zone of each wire type (CuNiTi or SmartArch™).
Figure 27: Comparison of bending curves of all zones of CuNiTi.

Figure 28: Comparison of bending curves of all zones of SmartArch™.
Figure 29: Representative bending curve of the central incisor zone of CuNiTi.

Figure 30: Representative bending curve of the lateral incisor zone of CuNiTi.
Figure 31: Representative bending curve of the canine zone of CuNiTi.

Figure 32: Representative bending curve of the 1st premolar zone of CuNiTi.
Figure 33: Representative bending curve of the 2nd premolar zone of CuNiTi.

Figure 34: Representative bending curve of the 1st molar zone of CuNiTi.
Figure 35: Representative bending curve of the 2nd molar zone of CuNiTi.

Figure 36: Representative bending curve of the central incisor zone of SmartArch™.
**Figure 37:** Representative bending curve of the lateral incisor zone of SmartArch™.

**Figure 38:** Representative bending curve of the canine zone of SmartArch™.
Figure 39: Representative bending curve of the 1\textsuperscript{st} premolar zone of SmartArch\textsuperscript{TM}.

Figure 40: Representative bending curve of the 2\textsuperscript{nd} premolar zone of SmartArch\textsuperscript{TM}.
Figure 41: Representative bending curve of the 1\textsuperscript{st} molar zone of SmartArch\textsuperscript{TM}.

Figure 42: Representative bending curve of the 2\textsuperscript{nd} molar zone of SmartArch\textsuperscript{TM}.
CHAPTER 5: DISCUSSION

The thermal and mechanical testing performed in this investigation is one of the first times Ormco’s SmartArch™ wires have been subject to independent quality control laboratory evaluation. Though multi-force archwires with three stiffness zones have proliferated the market over the past few years, the SmartArch™ wire has recently entered the market with a major differentiator—it is advertised to be preprogrammed with seven distinct force profiles for each tooth in the arch from the central incisor to the second molar. According to Roberts et al., these individual force levels were calculated by considering the average root and PDL surface area in addition to the interbracket distances (Roberts et al., 2019). Viecilli and Burstone performed finite element analysis to determine what the optimal PDL compressive stresses for average-sized teeth are, and these results were used to determine the specific SmartArch™ force zone profiles (Viecilli and Burstone, 2015; Olsen, 2020). Due to the more recent development of the SmartArch™ wires, there have not been any independent investigations to support the claims made by the manufacturers about the distinct force zones. Therefore, the aim of this study was to evaluate force distribution and phase transformation properties of this seven-force gradient nickel-titanium orthodontic archwire to understand the advertised force levels.

In this in vitro study, a differential scanning calorimeter (DSC) was used to assess the thermal properties of the SmartArch™ wire. These thermal values also serve as important indicators of the mechanical properties of the archwire. Though various parameters were tested through this thermal technique, one of the most important metrics to understanding the behavior of the wires is the austenite-finish (Aₐ) temperature which
represents the metal alloy’s transition temperature. According to Chybowski et al.,
greater $A_f$ temperatures correlate with smaller force values and vice versa (Chybowski,
2014). For example, a 27°C superelastic NiTi wire should have a greater force profile
than that of a 35°C superelastic NiTi wire, where the temperature value represents the
advertised $A_f$ temperature.

The current results indicated that almost all the thermal values (apart from cooling
enthalpy) between the SmartArch™ wires and CuNiTi wires along each zone were
significantly different from each other. These results seem reasonable because each
CuNiTi wire was uniformly heat treated to one $A_f$ temperature to deliver a single force
throughout the archwire while the SmartArch™ wires were laser conditioned to have
various $A_f$ temperatures and deliver various forces throughout a single archwire.
Therefore, it would make sense that the thermal parameters were significantly different
between the two wire types.

Additionally, after comparing the individual zones of the CuNiTi wires, the
results indicated that there were no significant differences between the zones for almost
all the thermal values (apart from cooling enthalpy and cooling endset temperature) as
seen in Table 1. This is also reasonable because the $A_f$ temperature should be consistent
throughout the single archwire and therefore should not show any significant differences.
Additionally, as expected, the individual zones for the SmartArch™ wire did show
significant differences between each other for all the thermal values as seen in Table 2.
These findings appear to reflect the variable thermal and mechanical profiles of the
SmartArch™ wire as described by the manufacturer.
However, when comparing the pattern of the SmartArch™ thermal values of each zone to the manufacturer’s data, there were some inconsistencies. Since the \( A_f \) temperature correlates with force values, the manufacturer’s force analysis can be compared to the measured DSC values. The two zones with the highest \( A_f \) were the lateral incisor (22.1°C) and the 1\textsuperscript{st} premolar (22.7°C) which corresponds with the lowest force values. While the manufacturer agrees that the lateral incisor zone should have the lowest force, the 1\textsuperscript{st} premolar zone is advertised as the third lowest. The next highest \( A_f \) values were the canine (21.4°C) and 2\textsuperscript{nd} premolar (21.4°C) zones. While the 2\textsuperscript{nd} premolar reflected the manufacturer’s data, the canine value was inconsistent with the claims as this zone was advertised as one of the highest forces in the arch after the molars. The central incisor \( A_f \) temperature (19.5°C) placed it as the fifth highest force and was the most inconsistent with the manufacturer’s claim that the central incisor zone should deliver the second lowest force. Lastly, the results suggest that the \( A_f \) temperature of the 1\textsuperscript{st} molar (18.3°C) and 2\textsuperscript{nd} molar (18.3°C) are the same and therefore should have the highest force profile in the arch which was consistent with the manufacturer’s claims. Therefore, it appears that there are distinct force zones throughout the SmartArch™ wire, but not in the pattern suggested by the manufacturer.

When comparing the \( A_f \) temperature of the 27°C CuNiTi wires to the manufacturer’s claims, there were some inconsistencies as well. The average \( A_f \) temperature of the wires was approximately 24°C while the manufacturer claims it to be 27°C. Therefore, the \( A_f \) of the 27°C CuNiTi wires in this study were within approximately 3°C of the manufacturer’s claims. The results of another study by Biermann et al. suggested that their results were within 2°C from the manufacturer’s
claims; however, the measured $A_f$ temperatures were in the 29°C range (Biermann et al., 2007). A possible reason for the differences between the current study, the Biermann et al. study, and the manufacturer’s findings may be attributed to the production lot of wires tested, the heat treatment manufacturing process, the exact metal composition of the wires, etc. which may have altered the transition temperatures of the wires.

In this *in vitro* study, a three-point bend test was used to assess the mechanical properties and load-deflection behavior of the SmartArch™ wire. Results indicated that almost all the bending values (17 of the 22 parameters) between the SmartArch™ wires and CuNiTi wires along each zone were significantly different from each other. These results were reflective of the manufacturer’s findings because each CuNiTi wire was uniformly heat treated to deliver a single force throughout the archwire while the SmartArch™ wires were segmentally laser conditioned to deliver various forces throughout a single archwire. Therefore, it is logical that the bending parameters were significantly different between the two wire types.

Additionally, after comparing the individual CuNiTi zones with each other, the results indicated that there were significant differences between the zones for all the bending parameters as displayed in Tables 3 and 4. This is inconsistent with the thermal analysis which showed a more uniform force profile. This is also inconsistent with the literature that states these single-force wires should not exhibit a force gradient as the current data appears to suggest. These differences in bending values between zones could be attributed to an arch-form shape factor effect where some wire segments that were tested had more curvature than others.
Additionally, as expected, the individual zones for the SmartArch™ wire did show significant differences between almost all bending parameters (apart from percent recovery) as seen in Tables 5 and 6. These findings appear to reflect the variable mechanical force profiles of the SmartArch™ wire as described by the manufacturer.

When comparing the pattern of the SmartArch™ bending values for each zone to the manufacturer’s data, there were some similarities, but the pattern of findings does not match the advertised values entirely. From the SmartArch™ manufacturer’s claims, the lateral incisor zone has the lowest force. After that, the lowest to highest advertised force zones are the central incisor, then the 1st and 2nd premolars, then the canine, and finally the 1st and 2nd molars which have the highest forces. When looking at the current study’s SmartArch™ bending data, with particular interest in the deactivation force at 3 mm of deflection, the biggest inconsistencies lie within the central incisor zone and the 1st and 2nd molars. The central incisor force value is the third highest value after the 1st and 2nd molars compared to the manufacturer’s claim that it has the second lowest force profile. Additionally, the manufacturer claims that the 1st and 2nd molars have similar force values with each other. However, the current findings suggest that the 1st molar zone has a significantly higher force value than the 2nd molar zone. Therefore, while it does appear that there are distinct force zones throughout the SmartArch™ wire, the force zone distribution is not in the pattern suggested by the manufacturer.

In summary, the findings from the SmartArch™ DSC thermal analysis and three-point bend test were consistent with each other. However, the pattern of the SmartArch™ force distribution is inconsistent with the force distribution that is presented by the manufacturer. Therefore, orthodontists must use their best judgement and set realistic
expectations when deciding to use these force gradient NiTi archwires. A possible reason for these differences between the current study and the manufacturer’s findings may be attributed to the production lot of wires tested, the laser conditioned manufacturing process, the exact metal composition of the wires, etc. which may have altered the force profile of the wires.

Since the three-point bend test was utilized to assess the mechanical properties of the SmartArch™ wires in a lab setting, one limitation is that the exact force profile determined in this study may not directly translate to clinical force values due to extraneous variables such as changes in bony support, anomalous tooth and/or root morphology, degree of malalignment, variations in interbracket distance, etc. However, the general force distribution pattern from the current study can serve as an important indicator of the SmartArch™ wire’s clinical performance.
CHAPTER 6: CONCLUSIONS

The following conclusions can be made from the current investigation:

1. The SmartArch™ thermal analysis showed significantly different austenite-finish temperatures among the tooth zones. Therefore, it appears thermal treatment via laser modification can alter the metallurgy and phase transformation properties of the SmartArch™ wires.

2. The SmartArch™ bending analysis showed significantly different force values throughout the single archwire. Therefore, it appears thermal treatment via laser modification can alter the mechanical properties of the SmartArch™ wires.

3. The CuNiTi thermal analysis did not show any significant differences among the tooth zones.

4. The CuNiTi bending analysis, however, did show some significant differences between the zones. This may be attributed to a shape factor effect as some segments of the archwire were more curved than others.

5. Though the SmartArch™ thermal and the bending results from the current study are consistent with each other, these values are inconsistent with the force zone distribution that is shown by the manufacturer.
REFERENCES


