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Copper, Aluminum and Nickel: A New Monocrystalline Orthodontic Alloy

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Copper, Aluminum and Nickel: A New Monocrystalline Orthodontic Alloy

by

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A Thesis submitted in partial satisfaction of
the requirements for the degree
Master of Science in Orthodontics and Dentofacial Orthopedics

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September 2014

Each person whose signature appears below certifies that this thesis in his opinion is adequate, in scope and quality, as a thesis for the degree of Master of Science.

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ABBREVIATIONS

AO	American Orthodontics©
ANOVA	Analysis of Variance
AUC	Area Under the Curve
β-Ti	Beta-Titanium
CuAlNi	Copper, Aluminum, and Nickel Alloy (Material Under Study)
CuNiTi	Copper Nickel Titanium
NiTi	Nickel Titanium
RMO	Rocky Mountain® Orthodontics
SS	Stainless Steel
TMA	Titanium Molybdenum Alloy

ABSTRACT OF THE THESIS

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Orthodontic Alloy

by

Mark Wierenga

Master of Science, Graduate Program in Orthodontics and Dentofacial Orthopedics
Loma Linda University, September 2014
Dr. V. Leroy Leggitt, Chairperson

Introduction: This study was designed to evaluate, via tensile and bend testing, the mechanical properties of a newly-developed monocrystalline orthodontic archwire comprised of a blend of copper, aluminum, and nickel (CuAlNi). **Methods:** The sample was comprised of three shape memory alloys; CuAlNi, copper nickel titanium (CuNiTi), and nickel titanium (NiTi); from various orthodontic manufacturers in both 0.018” round and 0.019” x 0.025” rectangular dimensions. Additional data was gathered for similarly sized stainless steel and beta-titanium archwires as a point of reference for drawing conclusions about the relative properties of the archwires. Measurements of loading and unloading forces were recorded in both tension and deflection testing. Repeated-measure ANOVA ($\alpha = 0.05$) was used to compare loading and unloading forces across wires and one-way ANOVA ($\alpha = 0.05$) was used to compare elastic moduli and hysteresis. To identify significant differences, Tukey post-hoc comparisons were performed. **Results:** The modulus of elasticity, deflection forces, and hysteresis profiles of CuAlNi were significantly different than the other superelastic wires tested. In all tests, CuAlNi had a statistically significant lower modulus of elasticity compared to the CuNiTi and NiTi wires ($P < 0.0001$). The CuAlNi wire exhibited significantly lower loading and unloading forces than any other wire tested. In round wire tensile tests, loading force at all deflections was significantly lower for CuAlNi than CuNiTi or NiTi ($P < 0.0001$). In tensile testing, the CuAlNi alloy was able to recover from a 7 mm extension (10% elongation) without permanent deformation and with little to no loss in force output. In large-deflection bend tests at 4, 5, and 6 mm deflection, CuAlNi showed the significantly lowest loading forces across the three wire materials ($P < 0.0001$). The NiTi wires showed up to 12 times the amount of energy loss due to hysteresis compared to CuAlNi. CuAlNi showed a hysteresis loss that was significantly less than any other wire tested in this study ($P < 0.0001$). **Conclusions:** The relatively constant force delivered for a long period of time during the deactivation of this wire, the minimal hysteresis loss, the low force output in deflection, and the relatively low modulus of elasticity suggest that CuAlNi wires should be considered an important material addition to orthodontic metallurgy.

CHAPTER ONE

REVIEW OF LITERATURE

Andreasen first introduced shape memory alloys to the field of orthodontics in the early 1970s.[1] Since then, shape memory alloys have been attractive for use as archwires with their superelastic properties in addition to shape memory mechanics. Since the 1970s, the formulation of these alloys has been adjusted to meet the demands of the orthodontist. The most recent development of a copper, aluminum and nickel monocrystalline hyperelastic archwire shows promise to continue the progression of improved clinical performance. As with any new wire, it is important to understand the composition of the wire and its mechanical properties in order to evaluate its potential clinical usefulness.

Nickel-titanium (NiTi) wire was originally developed during the 1960's by William Buehler. Through the efforts of Andreasen and Unitek in the early 1970's, the first NiTi alloy was marketed to orthodontists as Nitinol™, an acronym for nickel titanium and its origin at the Naval Ordnance Laboratory in Silver Springs, Maryland.[2,3] What was so attractive about this composition of nickel and titanium was its low spring-back force following activation. Compared to the other orthodontic archwires available, Nitinol delivered only one-fifth to one-sixth the force per unit of deactivation.[4] When Andreasen and Morrow analyzed Nitinol they reported a modulus of elasticity of 4.8×10^6 psi and an ultimate tensile strength of 230-300,000 PSI for Nitinol, compared with 28.5×10^6 psi and 280-300,000 PSI for the corresponding stainless steel tested.[2] Nitinol had a greater elastic limit, a very low modulus of elasticity, and possessed moderate strength. When comparing stainless steel and Nitinol wires of the same diameter, NiTi alloy wires delivered lower force levels while also displaying a significantly greater stored energy potential.[5] Since ideal archwires move teeth with light continuous forces,[6] this new addition to the orthodontists' arsenal was quickly adopted as an initial leveling archwire.

Multiple new combinations of nickel, titanium and other metals have been developed since the initial release of Nitinol that have unique properties while still maintaining the qualities of shape memory alloys. In 1985, Burstone introduced the orthodontic community to an austenitic NiTi developed in Beijing.[7] In this article, Burstone introduced this Chinese NiTi and compared it to the Nitinol and stainless steel archwires that were available at the time. He studied the wire's springback, stiffness, and the maximum moment using a flexural design study. Compared to the original Nitinol wire and a stainless steel wire of equal size, the Chinese NiTi had significantly lower stiffness, a larger springback, and a favorably lower maximum moment. The very next year, Miura, *et al*, published similar data about an austenitic NiTi developed in Japan. Miura used tensile and 3-point bend tests to make conclusions about the wire's unique properties compared to traditional nitinol alloy.[8]

Ormco developed a thermoelastic nitinol in 1994 that included copper in the traditional nickel and titanium alloy. Copper NiTi (CuNiTi) contains approximately 5-6% copper and small amounts of chromium. The addition of copper allows for a transformation between the softer, more pliable martensitic phase and the shape-retaining austenitic phase at different temperature ranges. Thus, these wires have the advantage of being body heat–activated, more easily engaged at room temperature, while transitioning into a more functional stiffness at temperatures encountered in the mouth.[9] The addition of copper has also been shown to reduce hysteresis - the energy lost in deformation.[10] This leads to a more stable delivery of force.[11] Adjusting the levels of chromium in the alloy changes the transformation temperature of the wire as the crystalline structure switches between martensite and austenite.[12]

Beta-titanium has been known to the clinician as the happy medium between NiTi and stainless steel since its introduction in the 1980's. Beta-titanium is marketed by the Ormco Corporation (Glendora, CA, USA) as titanium-molybdenum alloy (TMA). Beta-titanium is commonly produced at the ratio of 80% titanium, 11.5% molybdenum, 6% zirconium, and 4.5% tin.[13] Beta-titanium delivers lower biomechanical forces compared to stainless steel. The elastic modulus for beta-titanium wires is approximately 40% that of stainless steel and elgiloy blue wires. In addition to a lower elastic modulus, beta-titanium wires have significantly improved values of springback thus improving their working range for tooth movement.[14]

When classifying a new wire, its mechanical properties will assist in characterizing its clinical capabilities. A wire's modulus of elasticity is a basic material property that can reveal the relative stiffness of one wire to the next. The higher the modulus of elasticity, the greater the force magnitude delivered or stiffness of the wire.[6,15] As an archwire is bent, the outer curvature of the wire at the bend is placed under tensile forces while the wire at the inner portion of the bend is compressed. A wire that can withstand increased levels of tensile stress without permanent deformation will thus be able to return to its original shape after the bending force is released. For this reason, tensile and 3-point bend testing are valuable mechanical tests to compare different alloys used in orthodontics. Tensile testing was carried out soon after the initial development of Japanese NiTi. By measuring wire length before and after sequential elongation of an archwire, Miura, Mogi, Ohura, and Hamanaka were able to draw conclusions regarding the wire's behavior in an orthodontic environment.[16] Both tensile testing and the 3-point bending method are not directly transferable to the clinical setting, rather, they have been employed as physical property tests. These methods focus more on the physical and biomechanical properties of the wires, offer reproducibility, and are useful for purely theoretical evaluations. Both are standardized testing methods that make comparison to other studies possible. Unfortunately, there has not been a proper methodology developed that addresses the unusual properties found in superelastic wires. Most researchers have chosen to adhere to 3-point bend testing as described by the American Dental Association specification number 32.[17] This standard was originally formulated for stainless steel wires and was developed before the NiTi wire was introduced into mainstream orthodontic use. In many studies on superelastic wires, some investigators have developed their own testing methods in an attempt to quantify the bending characteristics of the wires beyond the traditional 3 mm of deflection. A consensus has not been reached on whether to continue using the existing 3-point bend test method, or to adopt a new and improved method of testing.

Another important characteristic of orthodontic archwires to analyze is their time-dependent properties and responses to repeated masticatory forces over time. Because an archwire remains in the patient's mouth for weeks to months at a time, it is important that the wire maintains its activity over time despite the continual cycling produced by repeated bends of the wire. In his article describing Chinese NiTi, Burstone investigated how a 6.5 mm vertical discrepancy would affect the deformation of NiTi compared to stainless steel and Nitinol. He compared the wires at time periods of one minute, one hour, and three days. After each time period, he analyzed and quantified the permanent deformations of the wire.[6] He found a favorable response to his austenitic NiTi over extended periods of deflection. Similar studies are needed to classify new orthodontic archwires.

In a system of brackets and wires, the effectiveness of the wire is directly correlated to the friction that exists with it and the brackets. Schumacher, Bourauel, and Drescher initiated a study of friction during the deactivation of leveling archwires. Their study showed a substantial decrease in the effective springback-force during deactivation due to friction elsewhere between the arch wire and brackets. Schumacher and colleagues found as much as a 50% reduction in the deactivation-force due to this friction.[18] For this reason, the friction values of any new orthodontic wire must be appropriately examined.

In a typical superelastic force deflection curve, there is a difference between the forces produced by a wire as its loaded compared to the unloading force produced. The areas of the curve showing nearly constant stress are the loading and unloading plateaus. The loading plateau represents the period during which the austenitic crystalline structure is stress-induced into martensite. As the load is removed, the stress-induced martensite transforms back into austenite along the unloading plateau. The loading plateau stress is always greater than the unloading plateau stress but the amount of difference is a key determinant of material properties. This difference in stress at loading and unloading plateaus is hysteresis in the system.[19] Bending the tines of a fork back and forth will demonstrate hysteresis as the metal becomes less responsive with repeated applications of force. With repeated bending, the metal builds up a lag in response to the same force. Hysteresis has been said to lead to unpredictable unloading forces potentially exceeding levels of patient comfort, resistance to sliding in brackets, and wires taking a permanent set or exhibiting incomplete recovery upon high straining.[13] Larger strains in the wire induce greater hysteresis for nickel titanium alloys, thus greater malocclusions are more likely to induce permanent wire deformation and a more unpredictable hysteresis loss. Studies have shown that the commercially available NiTi alloys behave in a variable manner, often deviating from superelasticity.[20] In a 2007 study by Bartzela, Senn, and Wichelhaus, 48 commercially available NiTi wires from five manufacturers were tested to determine if they were superelastic as advertised. In their study, they found that only 29 of the studied archwires (60%) showed true superelasticity. Of the remaining 19 archwires, seven were borderline superelastic, three were borderline nonsuperelastic, and three developed a permanent set after traditional three-point bend testing.[21]

In a polycrystalline wire, grains are separated by grain boundaries. It is at the grain boundaries where the grains slip past each other to result in a deformed wire. With the development of a monocrystalline wire, no grain boundaries exist. For that reason, repeatable and complete shape recovery has been obtained even at greater than 10% percent deformation. According to the manufacturer, this shape recovery correlates to three times greater than that of Nitinol.[22]

The Copper, Aluminum, and Nickel alloy (CuAlNi) has been referred to as being “hyperelastic.” When traditional superelastic archwires transform from one crystalline structure to another, energy is lost to hysteresis. In hyperelastic transformations, the energy is absorbed and released at nearly constant force, so that constant acceleration is attainable.[23] Because the range of strain recovery is so far beyond the maximum strain recovery of both conventional polycrystalline shape memory alloy materials and non-shape memory metals and alloys, such repeatable strain recovery properties of single crystal shape memory alloy has been referred to as hyperelastic.[24] To the orthodontist, this is especially favorable as the forces placed on the teeth must be of sufficient pressure to stimulate movement, but not enough to cause necrosis of the bony tissue or resorption of the roots. [25]

Hyperelastic alloys like the CuAlNi alloy under study are purported to have properties enabling them to undergo large recoverable distortions. The initial claims from its originators suggest it can withstand distortions at least an order of magnitude greater than that which could be obtained if the component were made of non-shape memory metals and alloys, and nearly an order of magnitude greater than can be obtained with polycrystalline shape memory alloy materials.[22, 23] Because the CuAlNi wire is monocrystalline, the hysteresis and the unloading curves are much more predictable than with a polycrystalline wire such as NiTi or CuNiTi. The study that follows was designed to quantify these mechanical properties of CuAlNi archwires in order to begin discovering its clinical usefulness.

CHAPTER TWO

COPPER, ALUMINUM AND NICKEL: A NEW MONOCRYSTALLINE ORTHODONTIC ALLOY

by

Mark Wierenga

Master of Science, Graduate Program in Orthodontics and Dentofacial Orthopedics Loma Linda University, September 2014
Dr. V. Leroy Leggitt, Chairperson

Abstract

Introduction: This study was designed to evaluate, via tensile and bend testing, the mechanical properties of a newly-developed monocrystalline orthodontic archwire comprised of a blend of copper, aluminum, and nickel (CuAlNi). **Methods:** The sample was comprised of three shape memory alloys; CuAlNi, copper nickel titanium (CuNiTi), and nickel titanium (NiTi); from various orthodontic manufacturers in both 0.018" round and 0.019" x 0.025" rectangular dimensions. Additional data was gathered for similarly sized stainless steel and beta-titanium archwires as a point of reference for drawing conclusions about the relative properties of the archwires. Measurements of loading and unloading forces were recorded in both tension and deflection testing. Repeated-measure ANOVA (α= 0.05) was used to compare loading and unloading forces across wires and one-way ANOVA (α= 0.05) was used to compare elastic moduli and hysteresis. To identify significant differences, Tukey post-hoc comparisons were performed. **Results:** The modulus of elasticity, deflection forces, and hysteresis profiles of CuAlNi were significantly different than the other superelastic wires tested. In all tests, CuAlNi had a statistically significant lower modulus of elasticity compared to the CuNiTi and NiTi wires ($P < 0.0001$). The CuAlNi wire exhibited significantly lower loading and unloading forces than any other wire tested. In round wire tensile tests, loading force at all deflections was significantly lower for CuAlNi than CuNiTi or NiTi ($P < 0.0001$). In tensile testing, the CuAlNi alloy was able to recover from a 7 mm extension (10% elongation) without permanent deformation and with little to no loss in force output. In large-deflection bend tests at 4, 5, and 6 mm deflection, CuAlNi showed the significantly lowest loading forces across the three wire materials ($P < 0.0001$). The NiTi wires showed up to 12 times the amount of energy loss due to hysteresis compared to CuAlNi. CuAlNi showed a hysteresis loss that was significantly less than any other wire tested in this study ($P < 0.0001$). **Conclusions:** The relatively constant force delivered for a long period of time during the deactivation of this wire, the minimal hysteresis loss, the low force output in deflection, and the relatively low modulus of elasticity suggest that CuAlNi wires should be considered an important material addition to orthodontic metallurgy.

Introduction

The field of orthodontics continues to develop with the introduction of new products. Orthodontic practitioners are constantly looking for more advantageous treatment protocols and technology. Research and technology in orthodontics are driven by the desire to decrease treatment time, costs and patient discomfort while increasing compliance and favorable health outcomes. In the past, as new orthodontic wires have been developed, appropriate laboratory and clinical studies have been run to determine whether the new materials are suitable for clinical use in orthodontic practice.

Nickel-titanium (NiTi) wire was originally developed during the 1960's by William Buehler. Through the efforts of Andreasen and Unitek in the early 1970's, the first NiTi alloy was marketed to orthodontists as Nitinol^{TM}, an acronym for nickel titanium and its origin at the Naval Ordnance Laboratory in Silver Springs, Maryland.[1,2,3] What was so attractive about this composition of nickel and titanium was its low spring-back force following activation. Compared to the other orthodontic archwires available, Nitinol delivered only one-fifth to one-sixth the force per unit of deactivation.[4] When comparing stainless steel and Nitinol wires of the same diameter, NiTi alloy wires delivered lower force levels while also displaying a significantly greater stored energy potential.[5] Since ideal archwires move teeth with light continuous forces,[6] this new addition to the orthodontists' arsenal was quickly adopted as an initial leveling archwire.

Multiple new combinations of nickel, titanium and other metals have been developed since the initial release of Nitinol that have unique properties while still maintaining the qualities of shape memory alloys. In 1985, Burstone *et al* introduced the orthodontic community to austenitic NiTi.[7] In this article, Burstone compared a new formulation of NiTi developed in Beijing to the Nitinol and stainless steel archwires that were available at the time. He studied the wire's springback, stiffness, and the maximum moment. Compared to the original Nitinol wire, the austenitic Chinese NiTi had significantly lower stiffness, a larger springback, and a favorably lower maximum moment. The very next year, Miura, *et al*, published similar data about an austenitic NiTi developed in Japan. Miura used tensile and 3-point bend tests to make conclusions about the wire's unique properties compared to traditional nitinol alloy.[8]

Ormco introduced a thermoelastic nitinol in 1994 that included copper in the traditional nickel and titanium alloy. Copper NiTi (CuNiTi) contains approximately 5-6% copper. The addition of copper allows for a transformation between the softer, more pliable martensitic phase and the shape-retaining austenitic phase at different temperature ranges. Thus, these wires have the advantage of being body heat activated, more easily engaged at room temperature, while transitioning into a more functional stiffness at temperatures encountered in the mouth.[9] The addition of copper also has been shown to reduce hysteresis - the energy lost in deformation.[10] This leads to a more stable delivery of force.[11]

The recent introduction of a monocrystalline copper, aluminum and nickel alloy for orthodontic use is intended to create more biologically compatible tooth movement and exert more predictable forces. This study was designed to quantify pertinent mechanical properties of CuAlNi archwires in order to begin discovering its clinical usefulness. The null hypothesis was that there is no difference between CuAlNi and similarly-sized superelastic wires when measuring the modulus of elasticity, deflection forces, and stress-induced hysteresis. Conversely, the alternative hypothesis is that there is a difference in the three measures between CuAlNi and the other wires tested. Data was obtained and compared for the CuAlNi wire and a sampling of other currently available superelastic orthodontic wires of similar size. Additional data was gathered for similarly sized stainless steel and beta-titanium archwires as a point of reference for drawing conclusions about the relative properties of the new archwire. This information should provide an initial set of foundational information for the orthodontic community about CuAlNi archwires and potentially serve as a reference for further study into this wire's use and clinical effectiveness.

Materials and Methods

Ten wire types were evaluated. Nickel titanium (NiTi), thermally activated copper nickel titanium (CuNiTi), beta-titanium (β-Ti), titanium molybdenum (TMA) and stainless steel (SS) archwires were selected at random from well-known orthodontic manufacturers (Table 1). Three separate tests were run; 3-point bend, tensile, and 6 mm deflection tests (Fig 1-4). Each test was comprised of ten wires of each wire type tested (n=10).

Table 1. Wire Types Evaluated

Wire Type	Manufacturer	Dimension (in)
SS	RMO	0.018
TMA	Ormco	0.018
NiTi	AO	0.018
CuNiTi	3M Unitek	0.018
*CuAlNi	Ormco	0.018
SS	Ormco	0.019 x 0.025
β-Ti	AO	0.019 x 0.025
NiTi	3M Unitek	0.019 x 0.025
CuNiTi	RMO	0.019 x 0.025
*CuAlNi	Ormco	0.019 x 0.025

All testing was performed with a 1-kN electromechanical load frame (Instron 5944, Norwood, MA). 3-point bend and tensile test were performed in accordance with the International Organization for Standardization (ISO) methods described in the American National Standard and American Dental Association (ANSI/ADA) specification number 32.[17]

Tensile testing of the wires was carried out on a 70 mm gauge wire stretched 7 mm for analysis of wire characteristics (10% elongation). The load frame crossheads were separated at a rate of 2 mm per minute. Temperature was regulated at 36±1°C (Fig 1).

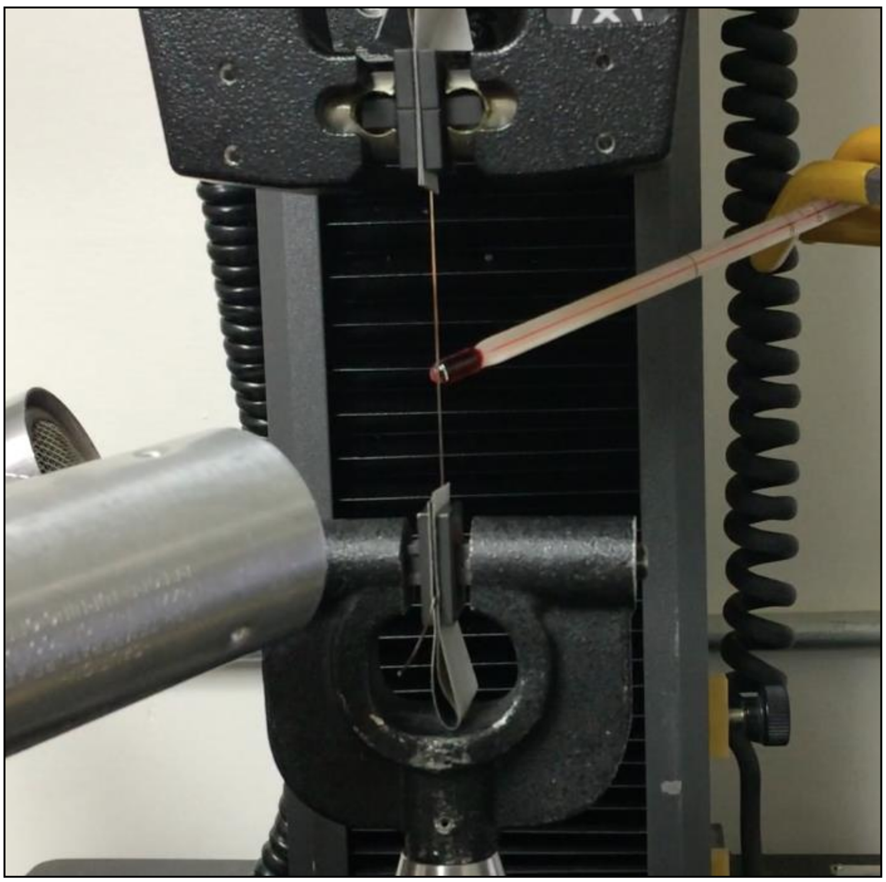

Fig 1. Tensile testing set-up

The 3-point bending test was carried out with wire at a length of 30 mm loaded at 36±1°C (Fig 2). A centrally placed indenter was used to deflect the wires 3.1 mm vertically across a 10 mm horizontal span at a rate of 7.5 mm per minute. A custom fabricated indenter and fulcrum were used both having radii of 0.10 mm in accordance with the ANSI/ADA specifications. Bending force was reported from the raw data at loading and unloading deflections of 0.5, 1.0, 2.0, and 3.0 mm.

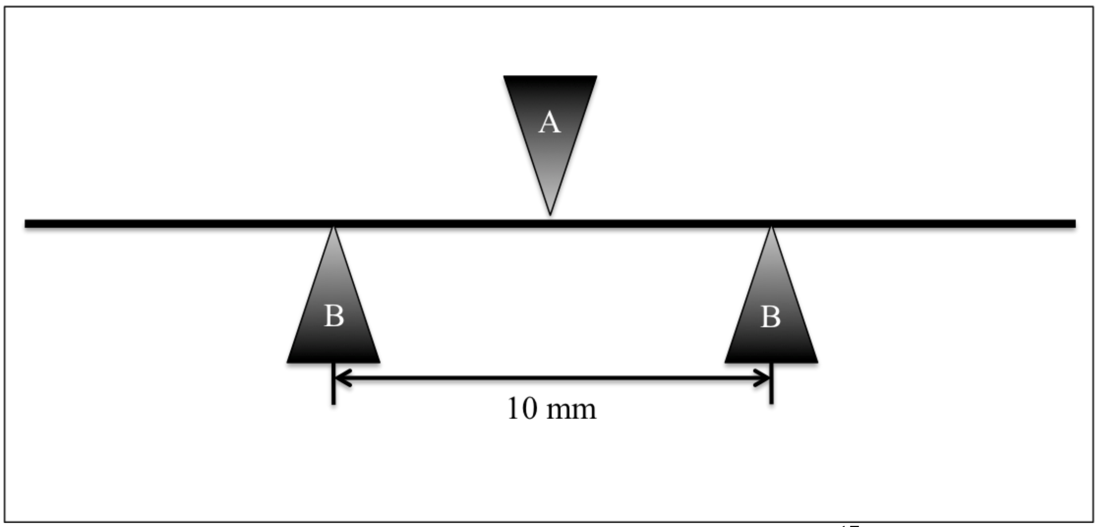

Fig 2. 3-point bend set-up. In accordance to ADA Specifications.[17] (A) Indenter (B) Fulcrum

To measure loading and unloading forces at deflections greater than established in the ADA specifications, a custom jig was constructed. Four Damon® self-ligating brackets (Ormco, Glendora, CA) were bonded to bovine enamel blocks adhered to a fixture as shown in Figure 3. Brackets were placed in positions representing a maxillary central, lateral, first premolar and second premolar with the load cell acting as a displaced canine. A 15.5 mm interbracket distance was used to simulate the average width between a lateral incisor and first premolar according to Moyers, et al[27] (Figs 3 and 4). Wire deflection was carried out gingivo-occlusally in the model to mimic intraoral conditions. Ten samples of each of the three 0.018” round superelastic wire types were deflected 6.1 mm across the 13.5 mm fulcrum distance using a centrally placed indenter with a crosshead rate of 7.5 mm/min at 36±1°C. Loading forces were measured and reported from the raw data at 4, 5, and 6 mm.

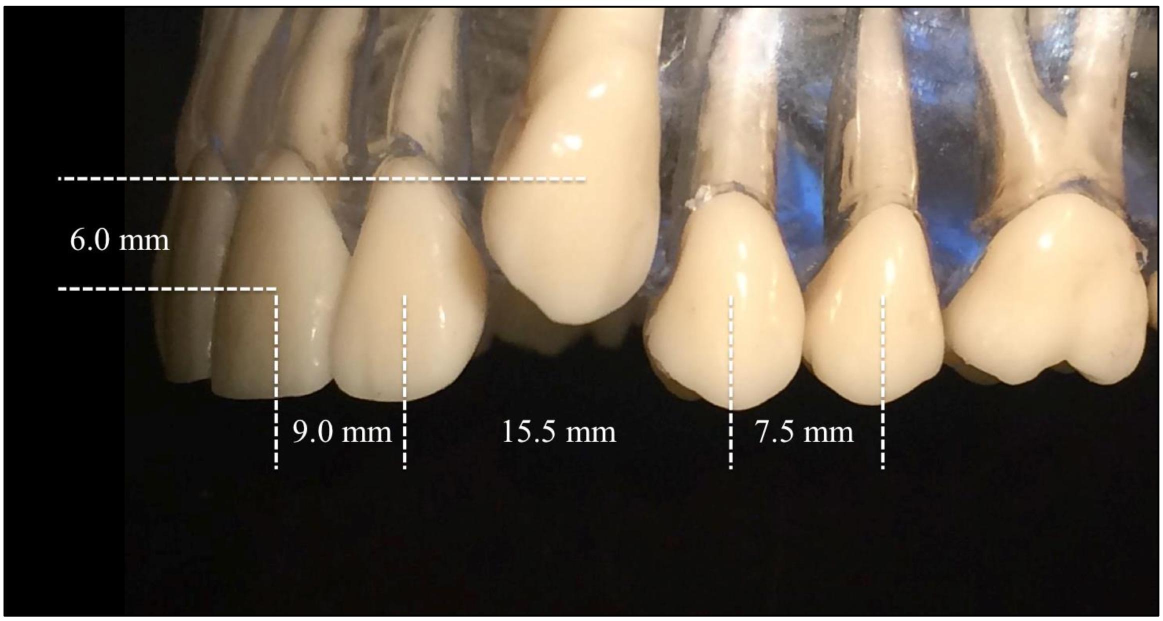

Fig 3. Mean interbracket distance diagram. Used as a reference for fabrication of large-deflection fixture (modified from Moyers et al[27]).

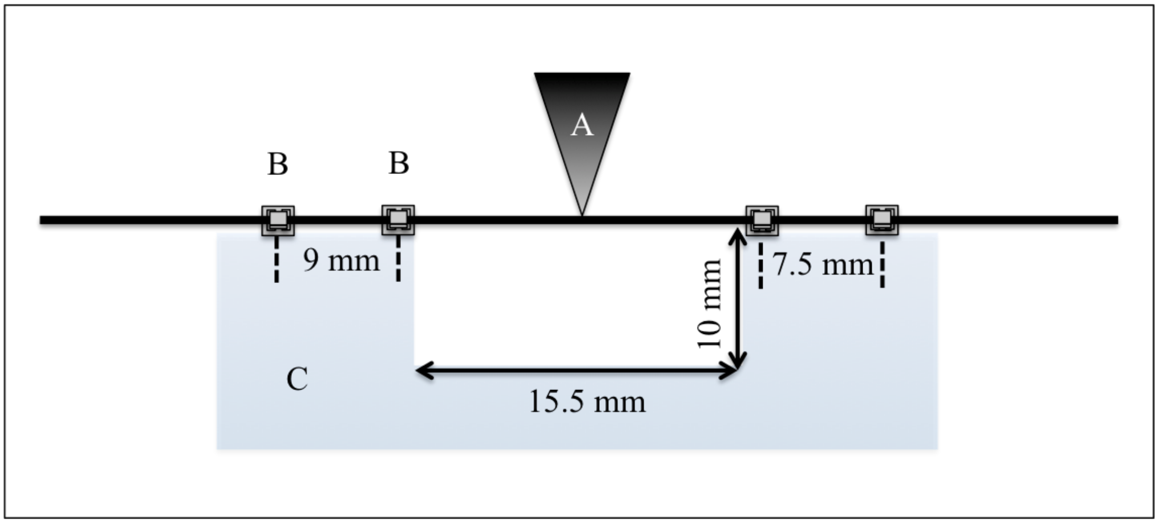

Fig 4. 6 mm deflection test set-up. (A) Indenter (B) Brackets (C) Custom Jig

Temperature regulation was carried out with the use of dual Varitemp Heat Guns (Master Appliance, Racine, WI) with a thermometer mounted directly adjacent to the wire. Temperature was set to 36±1°C to simulate intraoral temperatures and activate the wires accordingly. All loadframe testing data was measured and recorded with Bluehill 2 software (Instron, Norwood, MA).

Statistical Analysis

Statistical comparisons were performed with the Statistical Package for Social Sciences software for Windows (SPSS, Chicago, Ill). Repeated-measures ANOVA was performed to compare loading and unloading force across wires in all tensile and bending tests. The ANOVA model included wire material and compressive extension (mm) as main effects, as well as an interaction term between the two (wire × extension). One-way ANOVA was conducted to see if there were any significant differences in elastic modulus among the five wire materials. Comparison of hysteresis among the wire types was calculated by subtracting the area under the curve (AUC) of unloading forces from the AUC of loading forces. One-way ANOVA was used to compare mean hysteresis across the wires. Wherever significance was indicated, Tukey post-hoc comparisons were performed to reveal which of the wire materials were significantly different. The alpha level was set to be 0.05. ANOVA assumptions were verified with residual plots. No violation of ANOVA assumptions was present.

Results

Elastic Modulus

In all tests (3-point bend, tensile, and 6 mm deflection test) ANOVA showed there were significant differences. Tukey post-hoc testing identified the location of significant differences (Table 2, Appendix A).

CuAlNi had a statistically significant lower modulus of elasticity in all tests compared to the other four CuNiTi and NiTi wires (P <0.0001). In 3-point bend tests (round), all modulus values were all significantly different, except between the 3M CuNiTi and American NiTi (P= 0.51) archwires. The mean modulus of elasticity was significantly different between all rectangular wires in 3-point bend and tensile tests (P <0.0001). In tensile tests (round), there was no significant difference in modulus between 3M CuNiTi and American NiTi (P= 0.18). In the large-deflection tests, mean modulus values were all significantly different (P <0.0001). Stainless steel wires consistently demonstrated the highest modulus among all wires, with a mean modulus eight times greater than was measured for CuAlNi.

Table 2. Elastic modulus results (MPa) for all tests

3-point bend (round)

Wire	N	Mean	SD
CuAlNi	10	802.9	58.9
NiTi	10	2399.8	201.0
CuNiTi	10	2484.5	132.7
TMA	10	2667.9	64.1
SS	10	6818.9	75.1

One-way ANOVA: Main effect for Wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$, except between CuNiTi and NiTi ($P = 0.51$)

3-point bend (rectangular)

Wire	N	Mean	SD
CuAlNi	10	1278.1	54.9
NiTi	10	2794.8	167.0
CuNiTi	10	3589.0	251.3
β-Ti	10	4006.3	46.2
SS	10	9911.3	264.6

One-way ANOVA: Main effect for Wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$.

Tensile test (round)

Wire	N	Mean	SD
CuAlNi	10	14440.8	1386.5
NiTi	10	41722.1	2301.5
CuNiTi	10	43708.3	3239.2

One-way ANOVA: Main effect for Wire: P <0.0001

Tukey post-hoc tests show that CuAlNi has a significantly lower modulus than the other wires (P <0.0001)

Tensile test (rectangular)

Wire	N	Mean	SD
CuAlNi	10	8909.6	1127.7
CuNiTi	10	20031.8	3233.9
NiTi	10	24482.0	1881.4

One-way ANOVA: Main effect for Wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$.

6 mm Deflection (round)

Wire	N	Mean	SD
CuAlNi	10	587.8	43.5
CuNiTi	10	1479.0	78.7
NiTi	10	1747.2	45.4

One-way ANOVA: Main effect for Wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$.

Deflection Force

Repeated-measure ANOVA analyses were performed comparing loading and unloading forces across all wires in each of the three test types. Tukey post-hoc comparisons were used to see if loading and unloading forces were significantly different across wires at any extension or deflection (Appendix A). Means and SD of loading and unloading force by wire and extension are presented in Tables 3-7 including notations on the statistically significant differences.

In all tests (3-point bend, tensile, and 6 mm deflection tests), there were significant wire × extension interactions (all $P < 0.0001$). This indicated that the stress-strain curves were significantly different between wire types. In 3-point bend tests for both round and rectangular wires, loading forces at 0.5, 1, 2, and 3 mm of loading and unloading were significantly lower for CuAlNi than all other wire materials ($P < 0.0001$). In round wire tensile tests, loading force at all deflections was significantly lower for CuAlNi than CuNiTi or NiTi ($P < 0.0001$ for both). In rectangular wire tensile tests, loading forces at 2, 3, 4, 5, 6, and 7 mm were all significantly different across the three wire materials, with CuAlNi having the lowest deflection forces. Loading force was highest in NiTi wires, followed by CuNiTi and CuAlNi.

In 6 mm deflection tests, at 4, 5, and 6 mm extensions, CuAlNi showed the significantly lowest loading forces across the 3 wire materials. In unloading force, CuAlNi wires were significantly lower than CuNiTi or NiTi at 6 mm ($P < 0.0001$ for both). There was no significant difference between CuNiTi and NiTi ($P = 0.19$). There were no significant differences between the three wire types at 5 mm of unloading force due to binding in the brackets. At an unloading deflection of 4 mm, there was a statistically significant higher level of force for the CuAlNi as it was the only wire to have all 10 specimens avoid binding in the brackets following the 6 mm deflection. Of the 10 NiTi specimens run in this test, 7 became bound after the 6 mm deflection and of the 10 CuNiTi wires, 8 experienced binding after the 6 mm deflection (see Fig 13). These wires did not show permanent deformation after being removed from the apparatus post-test.

Table 3. Deflection forces (N) of round wire 3-point bend testing

	Extension (mm)	**CuAlNi**	**CuNiTi**	**NiTi**	**TMA**	**SS**	***P*-Value**
Load	0.5	$0.92\pm.06^{a}$	$2.47\pm.23^{b}$	$2.79\pm.23^{c}$	$2.64\pm.09^{d}$	$8.13\pm.30^{e}$	<0.0001
	1.0	$1.12\pm.11^{a}$	$3.05\pm.09^{b}$	$4.06\pm.14^{c}$	$5.60\pm.10^{d}$	$13.14\pm.17^{e}$	<0.0001
	2.0	$1.12\pm.10^{a}$	$3.38\pm.10^{b}$	$4.46\pm.14^{c}$	$6.75\pm.09^{d}$	$13.20\pm.27^{e}$	<0.0001
	3.0	$1.07\pm.09^{a}$	$3.35\pm.09^{b}$	$4.40\pm.13^{c}$	$6.21\pm.07^{d}$	$11.87\pm.14^{e}$	<0.0001
Unload	3.0	$0.78\pm.06^{a}$	$2.01\pm.08^{b}$	$2.94\pm.14^{\mathbf{c}}$	$2.80\pm.12^{\mathbf{c}}$	$4.89\pm.55^{d}$	<0.0001
	2.0	$0.72\pm.07^{a}$	$1.30\pm.10^{\mathbf{b}}$	$2.75\pm.15^{c}$	$1.75\pm.07^{d}$	$1.21\pm.10^{\mathbf{b}}$	<0.0001
	1.0	$0.71\pm.07^{a}$	$1.10\pm.08^{b}$	$2.35\pm.12^{c}$	$0.17\pm.01^{\mathbf{d}}$	$0.19\pm.01^{\mathbf{d}}$	<0.0001
	0.5	$0.79\pm.10^{a}$	$1.03\pm.09^{b}$	$2.07\pm.32^{c}$	$0.17\pm.01^{d}$	$0.19\pm.01^{e}$	<0.0001

^{a,b,c,d,e}: different letters denote statistically significant difference between forces (ANOVA at $\alpha = 0.05$)

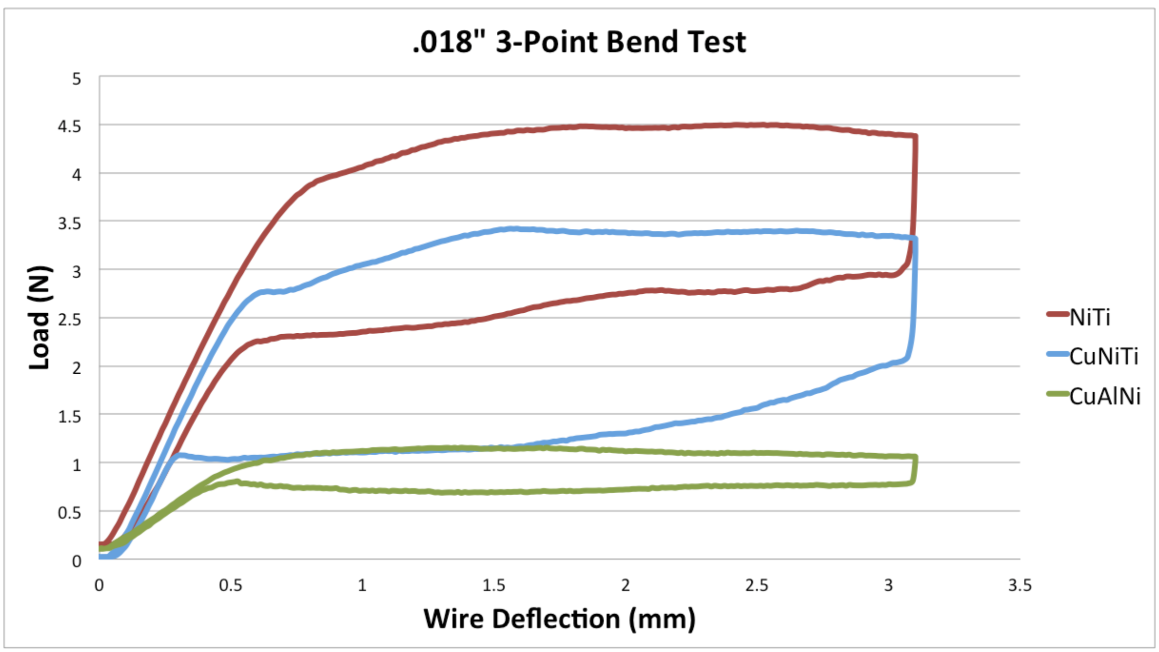

Fig 5. 0.018" 3-point bend test results (mean values of all specimens).

Table 4. Deflection forces (N) of rectangular wire 3-point bend testing

	Extension (mm)	**CuAlNi**	**CuNiTi**	**NiTi**	**β-Ti**	**SS**	***P*-Value**
Load	0.5	2.05±.10^{a}	5.07±.12^{b}	6.13±.35^{c}	8.60±.05^{d}	19.97±.48^{e}	<0.0001
	1.0	2.09±.13^{a}	5.67±.13^{b}	9.28±.39^{c}	15.24±.21^{d}	29.07±.25^{e}	<0.0001
	2.0	2.28±.12^{a}	6.76±.13^{b}	11.70±.39^{c}	18.87±.23^{d}	30.16±.32^{e}	<0.0001
	3.0	2.31±.13^{a}	7.18±.20^{b}	12.35±.38^{c}	17.34±.35^{d}	29.30±.68^{e}	<0.0001
Unload	3.0	1.32±.09^{a}	3.85±.13^{b}	6.44±.72^{c}	7.15±.64^{d}	16.36±.65^{e}	<0.0001
	2.0	1.30±.09^{a}	3.14±.13^{b}	6.30±.50^{c}	5.33±.14^{d}	0.02±.01^{e}	<0.0001
	1.0	1.40±.14^{a}	2.89±.15^{b}	5.48±.49^{c}	0.25±.02^{d}	0.02±.01^{d}	<0.0001
	0.5	1.46±.10^{a}	2.67±.16^{b}	4.39±.51^{c}	0.25±.02^{d}	0.02±.01^{e}	<0.0001

^{a,b,c,d,e}: different letters in rows denote statistically significant difference between forces (ANOVA at $\alpha = 0.05$)

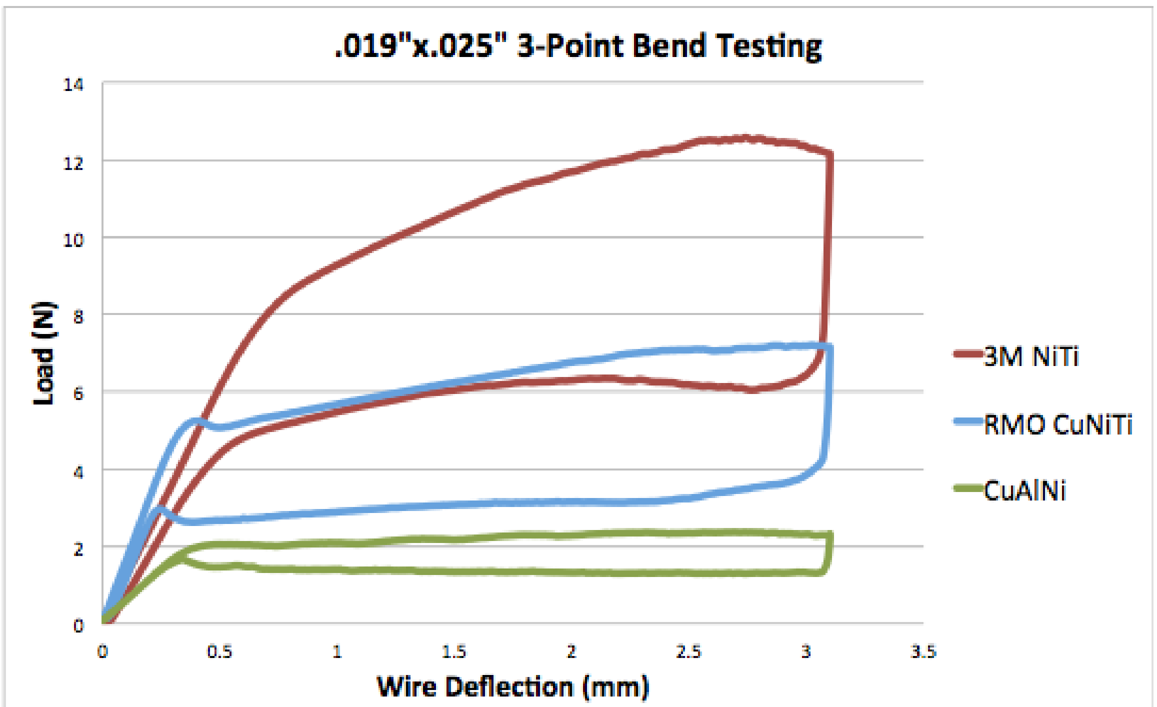

Fig 6. 0.019” x 0.025” 3-point bend test results (mean values of all specimens)

Table 5. Deflection forces (N) of round wire in tensile testing

	Extension (mm)	**CuAlNi**	**CuNiTi**	**NiTi**	***P*-Value**
Load	0.5	3.69±1.90^{a}	23.53±9.62^{b}	15.06±8.98^{b}	<0.0001
	1.0	18.08±2.80^{a}	61.25±3.50^{b}	58.08±9.19^{b}	<0.0001
	2.0	23.55±1.80^{a}	65.44±1.89^{b}	78.36±2.21^{c}	<0.0001
	3.0	24.12±1.89^{a}	67.86±1.75^{b}	81.09±2.49^{c}	<0.0001
	4.0	24.67±1.88^{a}	69.96±1.70^{b}	83.23±2.41^{c}	<0.0001
	5.0	24.82±1.41^{a}	74.08±3.00^{b}	84.86±2.17^{c}	<0.0001
	6.0	25.87±1.61^{a}	82.22±3.31^{b}	88.33±2.89^{b}	<0.0001
	7.0	27.27±1.52^{a}	104.24±6.50^{b}	105.67±4.39^{b}	<0.0001
Unload	7.0	27.27±1.52^{a}	104.246.50^{b}	105.65±4.39^{b}	<0.0001
	6.0	24.36±1.99^{a}	41.65±3.05^{b}	58.85±2.29^{c}	<0.0001
	5.0	22.68±1.61^{a}	28.27±2.60^{a}	46.82±2.41^{b}	<0.0001
	4.0	22.13±1.31^{a}	19.28±3.27^{a}	46.42±4.84^{b}	<0.0001
	3.0	21.55±1.44^{a}	20.19±1.91^{a}	44.09±2.42^{b}	<0.0001
	2.0	21.01±1.46^{a}	16.19±1.81^{a}	38.06±2.79^{b}	<0.0001
	1.0	12.41±3.02^{a}	0.67±1.67^{b}	17.76±4.08^{a}	<0.0001
	0.5	1.27±0.97	-0.91±0.15	0.64±0.61	0.20

^{a,b,c}: different letters denote statistically significant difference between load forces (ANOVA at $\alpha = 0.05$)

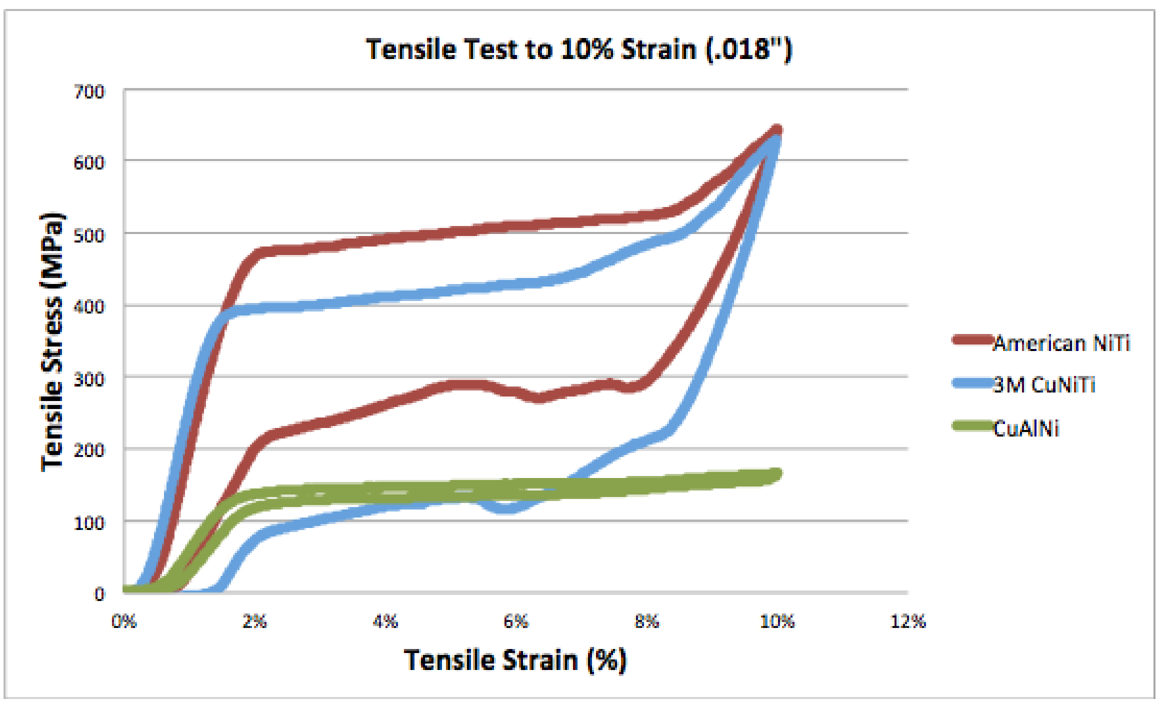

Fig 7. 0.018" tensile testing results (mean values of all specimens)

Table 6. Deflection forces (N) of rectangular wire in tensile testing

	Extension (mm)	**CuAlNi**	**CuNiTi**	**NiTi**	***P*-Value**
Load	0.5	2.18±1.17^{a}	5.93±6.73^{b}	10.68±7.49^{c}	<0.0001
	1.0	12.43±7.53^{a}	24.33±18.36^{a}	55.10±15.49^{b}	<0.0001
	2.0	35.50±10.88^{a}	86.16±8.37^{b}	139.58±9.71^{c}	<0.0001
	3.0	40.93±2.11^{a}	94.32±5.11^{b}	161.39±1.33^{c}	<0.0001
	4.0	42.07±1.85^{a}	101.42±5.84^{b}	170.06±2.51^{c}	<0.0001
	5.0	43.70±2.15^{a}	110.67±4.23^{b}	181.19±5.06^{c}	<0.0001
	6.0	45.08±2.06^{a}	120.00±6.13^{b}	210.54±10.56^{c}	<0.0001
	7.0	46.42±2.48^{a}	147.74±10.81^{b}	217.63±43.51^{c}	<0.0001
Unload	7.0	46.42±2.48^{a}	147.74±10.81^{b}	217.61±43.49^{c}	<0.0001
	6.0	40.84±2.25^{a}	70.51±5.17^{a}	140.11±17.20^{b}	<0.0001
	5.0	39.6±2.11^{a}	58.24±5.25^{a}	109.21±4.16^{b}	<0.0001
	4.0	38.76±2.12^{a}	50.65±6.83^{a}	95.82±11.50^{b}	<0.0001
	3.0	37.04±3.27^{a}	43.40±6.02^{a}	73.66±28.34^{b}	<0.0001
	2.0	29.45±12.42	16.03±5.51	29.32±23.46	0.04
	1.0	4.99±4.66	0.60±0.31	-0.73±1.09	0.60
	0.5	0.63±0.38	0.17±0.38	-1.5±.12	0.93

^{a,b,c}: different letters denote statistically significant difference between load forces (ANOVA at $\alpha = 0.05$)

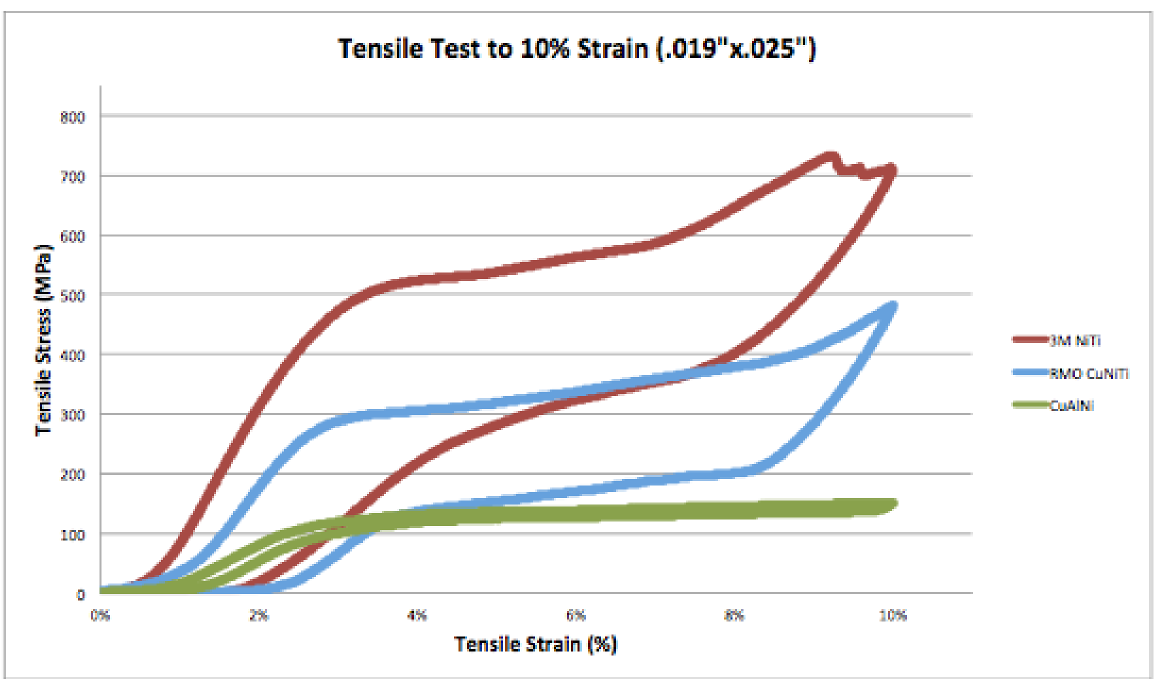

Fig 8. 0.019” x 0.025” tensile testing results (mean values of all specimens)

Table 7. Deflection forces (N) of round wire in bracket deflection testing

	Extension (mm)	**CuAlNi**	**CuNiTi**	**NiTi**	***P*-Value**
Load	4.0	1.55±.14[a]	6.92±.58[b]	8.28±.22[c]	<0.0001
	5.0	1.60±.16[a]	8.08±.82[b]	9.21±.31[b]	<0.0001
	6.0	1.66±.19[a]	9.24±1.11[b]	10.36±.45[b]	<0.0001
Unload	6.0	0.46±.05[a]	1.02±.26[b]	0.84±.16[b]	<0.0001
	5.0	0.17±.09	0.02±.04	0.06±.08	0.05
	4.0	0.30±.08	-0.01±.01	0.13±.22	<0.0001

[a,b,c]: different letters denote statistically significant difference between load forces (ANOVA at $\alpha = 0.05$)

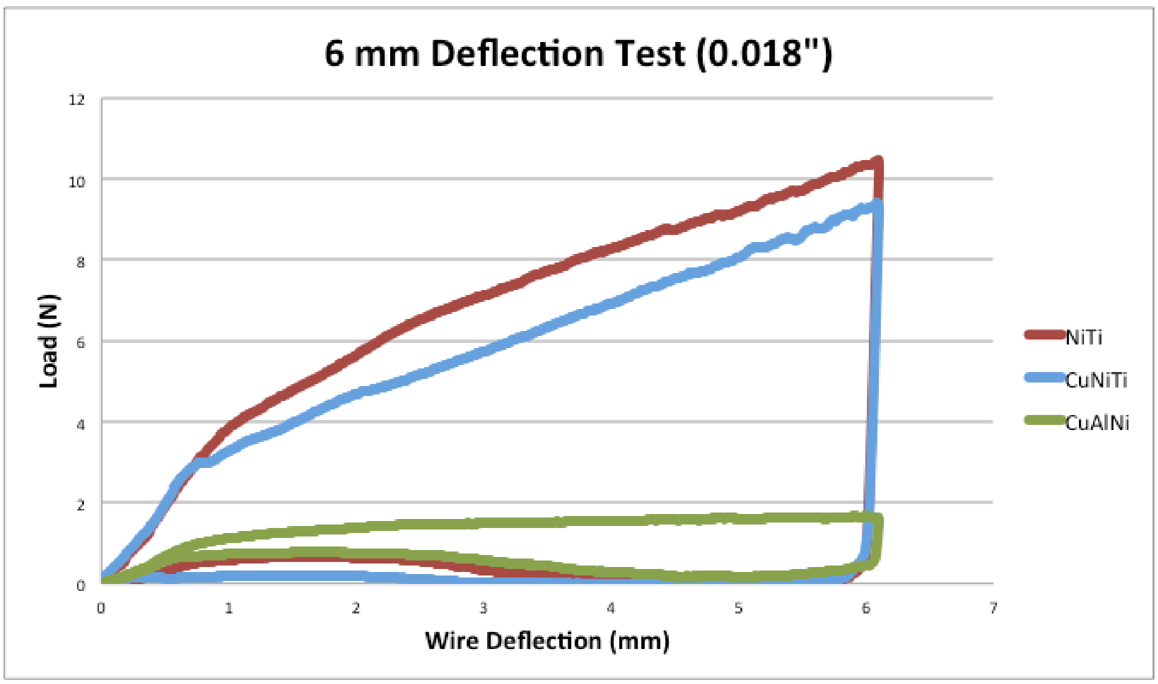

Fig 9. 0.018" 6 mm deflection testing results (mean values of all specimens)

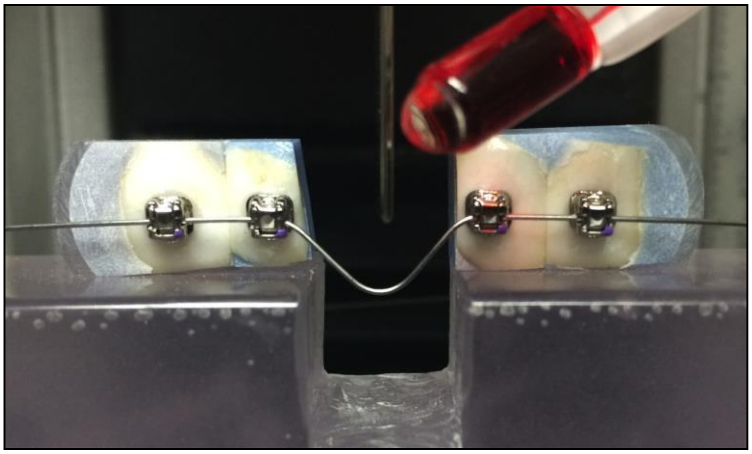

Fig 10. Wire binding after deflection (seen in 7 NiTi and 8 CuNiTi specimens)

Hysteresis

Hysteresis, the difference between loading and unloading force output, was calculated by subtracting the area under the curve (AUC) of unloading forces from the AUC of loading forces. One-way ANOVA was used to compare mean hysteresis across the wires. Means and SD of hysteresis by wire are presented in Table 8 including notations on the statistically significant differences.

CuAlNi showed a return force significantly closer to its displacement force compared to all other wires in all tests meaning CuAlNi wires had the lowest energy loss to hysteresis. The NiTi wires showed up to 12 times the amount of hysteresis loss compared to CuAlNi. In 3-point bend tests, stainless steel wires had the highest hysteresis, at approximately 30 times the magnitude of hysteresis loss of CuAlNi.

Table 8. Hysteresis results for all tests

3-point bend (round)

Wire	N	Mean	SD
CuAlNi	10	0.88	0.10
NiTi	10	3.90	0.08
CuNiTi	10	4.56	0.09
TMA	10	11.39	0.20
SS	10	27.17	0.37

One-way ANOVA: Main effect for wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$

3-point bend (rectangular)

Wire	N	Mean	SD
CuAlNi	10	2.12	0.22
CuNiTi	10	7.97	0.21
NiTi	10	11.64	0.78
β-Ti	10	31.96	0.84
SS	10	63.38	0.65

One-way ANOVA: Main effect for wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$

Tensile test (round)

Wire	N	Mean	SD
CuAlNi	10	25.7	2.8
NiTi	10	252.0	8.6
CuNiTi	10	322.1	9.6

One-way ANOVA: Main effect for wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$

Tensile test (rectangular)

Wire	N	Mean	SD
CuAlNi	10	48.0	4.0
CuNiTi	10	347.3	30.7
NiTi	10	549.6	67.1

One-way ANOVA: Main effect for wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$.

6 mm Deflection (round)

Wire	N	Mean	SD
CuAlNi	10	2.66	0.34
3M CuNiTi	10	15.63	1.69
NiTi	10	17.97	0.67

One-way ANOVA: Main effect for wire: P <0.0001

Tukey post-hoc tests show means are all significantly different at $\alpha = 0.05$.

Discussion

This study detected statistically significant differences in deflection forces, tensile responsiveness, and hysteresis profiles between the archwires. The CuAlNi wire exhibited statistically significant lower loading and unloading forces than any other wire tested. Data gathered in 3-point bend testing showed an average of 1.3 N of force in unloading for a 0.019”x 0.025” CuAlNi wire. This force level corresponds to published unloading forces of 3M Unitek’s 0.016” superelastic Nitinol wire (1.2 to 2.1 N at 3 mm deflection). [28] Therefore, CuAlNi may be more favorable in clinical situations requiring predictable, light force application as a larger cross-section behaves like a much smaller NiTi wire.

Of particular interest are the differences in loading and unloading forces seen between each of the wire types. When wires express a large stress hysteresis, forces can often exceed levels of patient comfort, resistance to sliding in brackets is dominated by binding forces between the bracket and wire, and wires may take a permanent set or exhibit incomplete recovery following high levels of strain. The 0.018” NiTi showed mean values of loading forces at 500 MPa while in unloading showed an average of 275 MPa. Thus the hysteresis loss in this wire approximated 45%. Similarly, the 0.018” 3M CuNiTi wire showed an average loading force of 425 MPa while in unloading averaged 150 MPa, a hysteresis loss of 65%. In contrast to the other wires, the 0.018” CuAlNi shows a nearly identical loading and unloading force with plateaus centering around 130 to 150 MPa, which remained consistent to 10% strain. Force levels for the CuAlNi returned to pretest values much more predictably than with NiTi and CuNiTi wires, both of which displayed permanent deformation and a return to 0 MPa before the crossheads returned to 0% strain. A similar response was seen with the rectangular 0.019” x 0.025” wires shown in Fig 6. As with the round wires, the NiTi shows higher force values in loading and unloading, with a return to 0% stress before the crossheads returned to their pretest position. Statistical analysis revealed a significant difference between the hysteresis profiles of all wires tested. The greater hysteresis for the NiTi wires represents a greater likelihood of wire fatigue. With repeated masticatory forces placed on wires throughout orthodontic treatment there is a cycling of loading and unloading stress plateaus on the wires, affecting the hysteresis and reducing force output of the wire.

The amount of crowding can be a relatively major consideration in the selection of an orthodontic archwire. With increased crowding, the CuAlNi wire is more likely to achieve complete adaptation into the bracket slot and perform with a lighter unloading force than previously possible with these large archwires. Yet while there may be statistically significant differences in the performance of individual wires in various mechanical test simulations, this does not necessarily indicate that such differences will exist in clinical performance. In a crowded dentition, high forces may be dissipated through interdental contacts as well as in overcoming friction amongst the brackets, wire, and ligatures.[29, 30, 31]

The obtained values for NiTi in the 3-point bend test at 3 mm deflection averaged 4.5 N while in the 6 mm deflection test the forces on loading at 3 mm deflection were 7 N. Just as is in clinical practice, friction between the wire and each of the four brackets is likely to have played a roll in the increased force values seen. It is likely that traditional 3-point bend tests underestimate the forces placed on teeth in loading as the more clinically-oriented 6 mm bend test shows a higher magnitude of force when friction of the brackets is taken into account. The opposite is true with unloading forces. With increased friction and binding of the archwires in the brackets, lower force values were produced by the wires as the 6 mm deflection was released. Yet, despite the attempt in the present study to design a model resembling clinical conditions, conclusions of the clinical performance of wires in the test must be made with caution.

The CuAlNi wire showed significantly lower force values in deflection than all other wires tested. It also produced nearly horizontal loading and unloading curves corresponding to a more consistent force delivery than any other wire tested. The clinical significance of consistent force delivery could mean lighter forces for patients treated with more crowding and therefore larger wire deflections. Caution must be taken in extrapolating numerical load values directly to clinical performance, yet the CuAlNi wire shows more consistent performance compared to traditional superelastic archwires. The concept of light forces producing more physiological and less painful tooth movement has been a matter of debate. While this study quantified the CuAlNi wire mechanically and showed that it produced a significantly lower level of force than NiTi and CuNiTi, additional laboratory and clinical research is needed to investigate the potential improvement in patient comfort with lighter forces delivered from this CuAlNi. Driving patient comfort as a priority in materials development will help improve not only the experience of our patients but also the practice environment of the orthodontic clinician.

Conclusions

1. According to the results of this study, CuAlNi shows a significantly lower modulus of elasticity compared to all other wires tested.
2. The CuAlNi alloy provided consistently lower force values in deflection and tension compared to all other wires studied.
3. The mechanical hysteresis loss of CuAlNi following deflection was significantly less than any other wire tested in this study.
4. Of the superelastic archwires tested, the NiTi wire provided the highest unloading values for every test deflection and model design.

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CHAPTER THREE

EXTENDED DISCUSSION

This study serves as an initial look into the new alloy's mechanical properties as they relate to orthodontics. While the study examined the wire according to ADA Specification Number 32, future studies are needed that more closely simulate the oral environment to better determine the CuAlNi wire’s clinical effectiveness. The extended deflection test run in this study showed a positive response from the CuAlNi wire in deflections of up to 6 mm over a 13.5 mm span. The extended capabilities of modern orthodontic archwires are more able to be shown in this test compared to frequently criticized and antiquated tests that have been run in the past. In future extended deflection tests, rather than wire indentation with a pointed fixture, a bracket could be used to grasp and deflect the wire. This would introduce additional friction components of multiple brackets as seen in a clinical setting.

Additional insight is needed to determine the sliding friction characteristics of the CuAlNi alloy. Having a highly flexible wire that produces consistent low force means it is a wire that will likely be used to unravel significant imbrication. In order to remain effective, the coefficient of sliding friction between the wire and bracket must be minimal. Due to CuAlNi being monocrystalline, the absence of grains and grain boundaries would likely lead to lower friction values. The manufacturing process and surface polishing of the CuAlNi must be optimized to ensure a smooth external surface. Reduced friction may have played a role in the superior performance of the CuAlNi wire which returned from the 6 mm deflection without binding. Confirmation of lower friction values of the CuAlNi wire compared to NiTi and CuNiTi wires will further develop the understanding of the capabilities of this new archwire.

Additional testing would be beneficial to determine the effects of mastication on the archwire and the response to the archwire over periods of weeks to months. Repeated cycling of the wire through its loading and unloading plateaus would provide additional insight into the potential effects of mastication on the archwires. A wire that shows greater hysteresis is likely to show a reduction in force output after repeated bending. In addition to repeated bending tests, time dependent studies that test the wire's responsiveness over extended periods of time would continue to develop our understanding of the capabilities of monocrystalline shape memory alloys.

Future studies are needed to fully understand the wire's clinical effectiveness and potential limitations. While lab studies are helpful to gain an initial understanding of new products, it is the clinical trials and case reports that will further the clinician's knowledge of this wire and increase the orthodontist's repertoire when it comes to aligning teeth.

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APPENDIX A

TUKEY POST-HOC ANALYSIS

3-Point Bend Test (round)

Extension (mm)	Wire	vs Wire	P-value Loading	P-value Unloading
0.5	CuNiTi	NiTi	0.001	<.0001
	CuNiTi	CuAlNi	<.0001	0.1307
	CuNiTi	TMA	0.5559	<.0001
	CuNiTi	SS	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	TMA	0.8184	<.0001
	NiTi	SS	<.0001	<.0001
	CuAlNi	TMA	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	TMA	SS	<.0001	0.9999
1.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	<.0001
	CuNiTi	TMA	<.0001	<.0001
	CuNiTi	SS	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	TMA	<.0001	<.0001
	NiTi	SS	<.0001	<.0001
	CuAlNi	TMA	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	TMA	SS	<.0001	0.9999
2.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	<.0001
	CuNiTi	TMA	<.0001	<.0001
	CuNiTi	SS	<.0001	0.9995
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	TMA	<.0001	<.0001
	NiTi	SS	<.0001	<.0001
	CuAlNi	TMA	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	TMA	SS	<.0001	<.0001
3.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	<.0001
	CuNiTi	TMA	<.0001	<.0001
	CuNiTi	SS	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	TMA	<.0001	0.9493
	NiTi	SS	<.0001	<.0001
	CuAlNi	TMA	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	TMA	SS	<.0001	<.0001

3-Point Bend Test (rectangular)

Extension (mm)	Wire	vs Wire	P-value Loading	P-value Unloading
0.5	NiTi	BetaTi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	SS	<.0001	<.0001
	NiTi	CuNiTi	<.0001	<.0001
	BetaTi	CuAlNi	<.0001	<.0001
	BetaTi	SS	<.0001	0.9938
	BetaTi	CuNiTi	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	CuAlNi	CuNiTi	<.0001	<.0001
	SS	CuNiTi	<.0001	<.0001
1.0	NiTi	BetaTi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	SS	<.0001	<.0001
	NiTi	CuNiTi	<.0001	<.0001
	BetaTi	CuAlNi	<.0001	<.0001
	BetaTi	SS	<.0001	0.9920
	BetaTi	CuNiTi	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	CuAlNi	CuNiTi	<.0001	<.0001
	SS	CuNiTi	<.0001	<.0001
2.0	NiTi	BetaTi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	SS	<.0001	<.0001
	NiTi	CuNiTi	<.0001	<.0001
	BetaTi	CuAlNi	<.0001	<.0001
	BetaTi	SS	<.0001	<.0001
	BetaTi	CuNiTi	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	CuAlNi	CuNiTi	<.0001	<.0001
	SS	CuNiTi	<.0001	<.0001
3.0	NiTi	BetaTi	<.0001	<0.001
	NiTi	CuAlNi	<.0001	<.0001
	NiTi	SS	<.0001	<.0001
	NiTi	CuNiTi	<.0001	<.0001
	BetaTi	CuAlNi	<.0001	<.0001
	BetaTi	SS	<.0001	<.0001
	BetaTi	CuNiTi	<.0001	<.0001
	CuAlNi	SS	<.0001	<.0001
	CuAlNi	CuNiTi	<.0001	<.0001
	SS	CuNiTi	<.0001	<.0001

Tensile Test (round)

Extension (mm)	Wire	vs Wire	P-value Loading	P-value Unloading
0.5	CuNiTi	NiTi	0.0024	0.9999
	CuNiTi	CuAlNi	<.0001	0.9877
	NiTi	CuAlNi	<.0001	1
1.0	CuNiTi	NiTi	0.9911	<.0001
	CuNiTi	CuAlNi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	0.0066
2.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	0.0298
	NiTi	CuAlNi	<.0001	<.0001
3.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	0.9999
	NiTi	CuAlNi	<.0001	<.0001
4.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	0.8235
	NiTi	CuAlNi	<.0001	<.0001
5.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	0.0031
	NiTi	CuAlNi	<.0001	<.0001
6.0	CuNiTi	NiTi	0.1663	<.0001
	CuNiTi	CuAlNi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001
7.0	CuNiTi	NiTi	0.9999	0.9999
	CuNiTi	CuAlNi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001

Tensile Test (rectangular)

Extension (mm)	Wire	vs Wire	P-value Loading	P-value Unloading
0.5	NiTi	CuAlNi	<.0001	0.9999
	NiTi	CuNiTi	<.0001	0.9999
	CuAlNi	CuNiTi	<.0001	0.9999
1.0	NiTi	CuAlNi	<.0001	0.9999
	NiTi	CuNiTi	<.0001	0.9999
	CuAlNi	CuNiTi	0.8308	0.9999
2.0	NiTi	CuAlNi	<.0001	0.9999
	NiTi	CuNiTi	<.0001	0.8386
	CuAlNi	CuNiTi	<.0001	0.8263
3.0	NiTi	CuAlNi	<.0001	<.0001
	NiTi	CuNiTi	<.0001	0.0002
	CuAlNi	CuNiTi	<.0001	0.9999
4.0	NiTi	CuAlNi	<.0001	<.0001
	NiTi	CuNiTi	<.0001	<.0001
	CuAlNi	CuNiTi	<.0001	0.9386
5.0	NiTi	CuAlNi	<.0001	<.0001
	NiTi	CuNiTi	<.0001	<.0001
	CuAlNi	CuNiTi	<.0001	0.2168
6.0	NiTi	CuAlNi	<.0001	<.0001
	NiTi	CuNiTi	<.0001	<.0001
	CuAlNi	CuNiTi	<.0001	0.0003
7.0	CuNiTi	NiTi	<.0001	<.0001
	CuNiTi	CuAlNi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001

6 mm Deflection Test

Extension (mm)	Wire	vs Wire	P-value Loading	P-value Unloading
4.0	CuNiTi	NiTi	<.0001	0.4245
	CuNiTi	CuAlNi	<.0001	0.0003
	NiTi	CuAlNi	<.0001	0.1849
5.0	CuNiTi	NiTi	0.0008	0.9999
	CuNiTi	CuAlNi	<.0001	0.3348
	NiTi	CuAlNi	<.0001	0.6467
6.0	CuNiTi	NiTi	0.0009	0.2401
	CuNiTi	CuAlNi	<.0001	<.0001
	NiTi	CuAlNi	<.0001	<.0001

APPENDIX B

FIVE WIRE TYPES IN 3-POINT BEND TEST