Investigation Of Force Decay In Esthetic Composite Orthodontic Archwires

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INVESTIGATION OF FORCE DECAY IN ESTHETIC COMPOSITE ORTHODONTIC ARCHWIRES

by

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ABSTRACT
INVESTIGATION OF FORCE DECAY IN ESTHETIC COMPOSITE ORTHODONTIC ARCHWIRES

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Introduction: Fiber-reinforced composite archwires have been developed to increase the esthetics of orthodontic appliances. Because polymer containing composites typically exhibit time-dependent stress-strain behavior, deflected fiber-reinforced composite archwires may experience a decrease in force over time. The goal of this research was to determine if esthetic fiber-reinforced composite archwires can maintain continuous light forces without undergoing extreme amounts of force decay.

Materials and Methods: Force decay was evaluated by comparing results of 3-point bending tests of nickel-titanium (NiTi) and fiber-reinforced composite archwires. Due to the impracticality of measuring force decay of a single archwire for 30 days, the following protocol was used: wire segments were tested in 3-point bending using a universal testing machine to a maximum deflection of 3.1 mm; next, each segment was placed in a custom-made jig designed to deflect each segment either 1 or 2 mm for 30 days. Each segment was once again tested in 3-point bending to examine consistency of the bending profile. Paired t-tests were used to statistically compare pre- and post-deflection forces. A control group consisting of wires not subject to the 30 day constant deflection was tested to ensure that the initial testing did not alter the second 3-point bend test.

Results: Statistically significant ($p<0.05$) differences in the pre- and post-deflection force delivery were evident in the BioMers 2 mm deflection group and all of the NiTi groups. The BioMers 2 mm deflection group failed to deliver consistent forces as the majority of the wires experienced crazing during the 30 day deflection period. Though there is a statistically significant difference found in each NiTi group, the decrease in force delivery is not clinically significant. This statistical difference may be attributed to the small standard deviations in the NiTi groups.

Conclusions: The BioMers 1 mm deflection group demonstrated that fiber-reinforced composite archwires are able to deliver a consistent force after 30 days of deflection. However, the clinical applicability of these fiber-reinforced composite archwires may be limited as they are unable to sustain deflections of 2 mm without experiencing crazing and loss of force delivery.

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CHAPTER 1
INTRODUCTION

Orthodontics, the first dental specialty, primarily deals with guidance, modification, correction and maintenance of the dento-facial complex. An orthodontist's treatment goals are to achieve a functional, esthetic and stable dental occlusion and simultaneously maintain or improve facial harmony and balance. However, patients are typically most concerned with esthetics, both during and after treatment (Huang et al., 2003). Currently, the most commonly used orthodontic appliances mainly consist of metal alloy braces made from stainless steel, cobalt-chromium alloy or titanium alloy (Huang et al., 2003). Orthodontic archwires are typically manufactured with 18-8 stainless steel, chrome-cobalt-nickel (Elgiloy), or titanium alloys (Valiathan & Dhar, 2006) such as nickel-titanium (NiTi) and beta-titanium (TMA). The appearance of these metal braces and wires on the teeth is very noticeable and considered by many potential patients to be unesthetic and undesirable. In recent years there has been an increasing focus on dental esthetics and the need for orthodontic treatment (Walton et al., 2010), which has led to an increase in adults seeking orthodontic treatment. As the number of adults seeking orthodontic treatment has increased, so has the demand for a more esthetic orthodontic appliance (Jeremiah et al., 2011). Clear tray aligners, lingual braces, ceramic braces and polymer braces are examples of the various esthetic orthodontic appliances currently available. Polymer and ceramic braces are more commonly used because clear tray aligners have many treatment limitations (Rosvall et al., 2009) and lingual braces require unconventional mechanics, have a high lab fee and may affect speech.

There have been many advances in the physical properties of the current alloy archwires, however they have mostly remained unesthetic. The use of an esthetic orthodontic archwire in concert with an esthetic bracket, which is not yet common place in orthodontics, is likely the next step to enhance the esthetics of orthodontic appliances. Alloy archwires coated with a tooth colored polymer have been developed for use during the initial treatment period (Rosvall et al., 2009). One recent attempt to achieve the desired appliance esthetics has been the development of a translucent fiber-reinforced composite archwire with properties similar to those of the ideal alloy archwire (Zufall & Kusy, 2000).

Fiber-reinforced composite materials have been discussed in the dental literature since the early 1960s (Valiathan & Dhar, 2006). They have had a variety of dental applications such as provisional bridges, retainers, space maintainers (Goldberg & Burstone, 1992; Jancar et al., 1994), endodontic posts and cores, fixed partial dentures, periodontal splints, orthodontic splints, and trauma stabilization (Cacciafesta et al., 2008). Fiber-reinforced composites have been used orthodontically as anchorage units (Burstone & Kuhlberg, 2000; Cacciafesta et al., 2005) and are now being developed for use as orthodontic archwires (Valiathan & Dhar, 2006). In addition to the esthetic concerns, fiber-reinforced composite archwires used in harmony with polymer or ceramic brackets would help to eliminate the allergenic potential of a nickel containing appliance (Valiathan & Dhar, 2006).

Because polymer containing composites typically exhibit viscoelastic or time-dependent stress-strain behavior, it is possible that a deflected fiber-reinforced composite archwire would experience a decrease in force over time. This potential for decrease in the amount of springback in the wire would result in less force available for tooth movement and decreased treatment efficiency (Zufall & Kusy, 2000).

The goal of this research was to determine if these esthetic fiber-reinforced composite archwires can maintain continuous light forces without undergoing extreme amounts of force decay. This study directly compared the amount of force decay exhibited by fiber-reinforced composite archwires from BioMers Products, LLC (Jacksonville, FL) to that of conventional nickel-titanium archwires (Nitinol Classic from 3M Unitek, Monrovia, CA).

CHAPTER 2
LITERATURE REVIEW

Orthodontic tooth movement is achieved through the application of prolonged pressure to a tooth which results in a biologic response of bone remodeling and tooth movement (Proffit et al., 2012). This force application is typically produced by engaging an orthodontic archwire into a bracket resulting in an elastically deformed wire that transfers its energy to the tooth during deactivation (Valiathan & Dhar 2006). For years the most commonly used materials for orthodontic archwires have been stainless steel, nickel-titanium, beta-titanium and cobalt-chromium alloys. More recently, efforts have been made to research and develop fiber-reinforced composite archwires suitable for use in clinical orthodontics (Cacciafesta et al., 2008). The most efficient and desirable form of tooth movement is produced through application of continuous light forces (Proffit et al., 2012). In order to achieve optimal force levels over time, it is best to use an archwire with ideal physical properties. Although there is not one material best suited for all stages of treatment, the ideal orthodontic archwire should have high strength, high formability, high resiliency, high springback, low stiffness, low friction and the ability to be soldered or welded. It should also be cost efficient, biocompatible and esthetic. (Kusy, 1997; Proffit et al., 2012; Valiathan & Dhar 2006).

History of Orthodontic Archwires

In 1887, Edward Angle developed the arch bow appliance, which is now considered to be the precursor of the orthodontic archwire. The arch bow, also known as the E-arch, utilized 0.032 to 0.036 inch round wires made of precious metals such as nickel-silver and platinum-gold alloy. The arch bow was threaded at its ends and was affixed to bands on the terminal molars. By utilizing a nut placed mesial or distal to the molar tube, this appliance could be activated to facilitate anteroposterior or transverse expansion to provide room for the malposed teeth which were individually ligated to the arch bow. Due to the size and stiffness of the arch bow, individual tooth movements and leveling of the arch were not possible (Nikolai, 1997; Proffit et al., 2012). To overcome the limitations of the E-arch, Angle began placing bands on each tooth. Each band was outfitted with a vertically positioned rectangular slot behind the tube. A ribbon arch of 0.010 x 0.020 inch gold wire was inserted into each slot and affixed with pins. The springiness of the ribbon arch allowed it to be successful at aligning the crowns of teeth, but unfortunately the appliance was unable to generate moments necessary for proper root position. In his quest to achieve mechanical control in all three planes of space, Angle developed the edgewise appliance. In the edgewise appliance the archwire slot was reoriented from vertical to horizontal, thus allowing the insertion of a continuous rectangular archwire from one side of the arch to the other (Proffit et al., 2012). The egdewise appliance, with a slot size of 0.022 x 0.028 inch, effectively defined the transition from the arch bow to the archwire. Precious metal alloys were initially used as the archwires for the edgewise appliance, but they lacked the stiffness and rigidity in such small cross-sections to provide the stabilizing procedures necessary in orthodontic treatment (Nikolai, 1997).

Stainless Steel Wires

Stainless steel was introduced as an orthodontic archwire material in 1929. When compared to precious metals it offered greater strength, higher modulus of elasticity, good corrosion resistance, and lower costs (Kapila & Sachdeva, 1989). With the advent of stainless steel, costly precious metals such as gold, silver and platinum alloys began to disappear from orthodontic appliances (Nikolai, 1997). Stainless steel is generally composed (all compositions will be given in wt%, unless noted) of 17-25% chromium, 8-25% nickel and less than 0.20% carbon, with the remainder being iron. A common formulation of stainless steel for orthodontic use is 18% chromium and 8% nickel, thus it is often referred to as 18-8 stainless steel (Proffit et al., 2012). Having the chromium content higher than 10-13% allows for the formation of an oxide layer which provides passivity to the wire, rendering the alloy “stainless” by increasing its corrosion resistance compared to plain carbon steel. Nickel content of at least 8% stabilizes the austenite structure and also improves the overall resistance to corrosion (Kusy, 1997). As the marketing and use of stainless steel in orthodontics increased, the use of gold was essentially abandoned (Kusy, 2002). Stainless steel archwires provide many beneficial treatment capabilities that were not previously available with precious metals. However, stainless steel archwires exhibit force levels higher than ideal with a low amount of springback (Valiathan & Dhar, 2006).

Cobalt-chromium Wires

Elgin Watch Company developed a complex alloy for their watch springs consisting of 40% cobalt, 20% chromium, 16% iron, 15% nickel, 7% molybdenum, 2% manganese, 0.14% carbon and 0.04% beryllium (Kusy, 1997; O'Brien, 2008). This cobalt-chromium alloy was later introduced to orthodontics as Elgiloy in the 1950s by Rocky Mountain Orthodontics. The stiffness of Elgiloy is similar to that of stainless steel with the added benefit of altering the strength and formability through heat treatment (Kusy, 1997). Heat treatment causes precipitation hardening of the alloy which results in increased resistance to deformation. The ideal temperature for heat treatment is 900°F (482°C) for 7 to 12 minutes in a dental furnace (Kapila & Sachdeva, 1989). Variable strength and resilience with consistent stiffness was made possible as Elgiloy archwires were eventually manufactured in four different resiliencies: soft (blue), ductile (yellow), semi-resilient (green), and resilient (red) in order of increasing resilience (Kapila & Sachdeva, 1989; Kusy, 1997). Blue Elgiloy can be easily manipulated and is recommended for use when considerable bending, soldering or welding is necessary. Most mechanical properties of cobalt-chromium wires are similar to those of stainless steel, however, cobalt-chromium wires exhibit longer function as a resilient spring and offer greater resistance to fatigue and distortion (Kapila & Sachdeva, 1989).

Nickel-titanium Wires

In 1962, the Navy developed a nickel-titanium alloy, named Nitinol as an acronym for nickel-titanium Naval Ordinance Laboratory (Kusy, 2002). Nitinol was found to exhibit a shape memory effect that allowed it to be deformed, clamped, heated and cooled into a specific shape that the wire was able to return to following additional deformations (Kusy, 1997). Superelasticity is an additional unique property of nickel-titanium alloys (Proffit et al., 2012). Recognizing the potential clinical benefits of shape memory and superelastic qualities, Dr. George Andreasen made strides through the University of Iowa and Unitek Company to bring this 50 at% nickel and 50 at% titanium alloy to orthodontics in 1974 (Kusy, 2002). Nickel-titanium alloy archwires, produced commercially by many different manufacturers, are available as NiTi, Nitinol, Orthonol, Sentinol and Titanal, among other names (Kapila & Sachdeva, 1989).

Nickel-titanium alloys can exist in more than one form or crystal structure; the martensite form exists at lower temperatures and the austenite form at higher temperatures. Shape memory and superelastic properties are related to the phase transitions within the nickel-titanium alloy. The transitional temperature at which phase transformation occurs for most alloys is typically hundreds of degrees. Fortunately, nickel-titanium alloys transform between martensite and austenite forms at lower temperatures. The initial nickel-titanium wires were unable to take advantage of the phase transformation effects as they were stabilized in the martensitic form. In the late 1980s, active austenitic nickel-titanium archwires exhibiting superelasticity were introduced. The benefit of the superelastic nickel-titanium archwires is that they deliver a relatively continuous light force whether they are deflected a small or large distance. This unique ability to deliver the same amount of force regardless of the degree of activation is due to the stress-induced phase transformation from austenite to martensite (Proffit et al., 2012).

Mechanical properties of nickel-titanium alloys, such as high springback, high flexibility and low modulus of elasticity make it beneficial for use as an initial orthodontic archwire (Kapila & Sachdeva, 1989). Titanium alloys exhibit high corrosion resistance due to the spontaneous formation of a titanium dioxide layer (Heakal & Awad, 2011). The high springback and large recoverable energy exhibited by nickel-titanium wires results in an increase in clinical efficiency as they allow for fewer archwire changes or activations and more constant force delivery. One distinct advantage of nickel-titanium wires is the ability to insert a rectangular archwire relatively early in treatment, which accomplishes simultaneous leveling, aligning and root positioning. The drawbacks of nickel-titanium alloys are that it has poor formability, a higher coefficient of friction than stainless steel and it cannot be welded or soldered (Kapila & Sachdeva, 1989).

Beta-titanium Wires

In 1977 an orthodontic wire was developed whereby the beta-phase of titanium was stabilized to room temperature which enabled the production of a corrosion resistant alloy with high springback, good formability and the ability to be welded (Kapila & Sachdeva, 1989; Kusy, 2002). Beta-titanium alloy is composed of approximately 80% titanium, 11.5% molybdenum, 6% zirconium and 4.5% tin (Kusy, 1997). Beta-titanium alloy is more commonly known as TMA, which is an acronym for titanium-molybdenum alloy. The properties of beta-titanium are somewhat intermediate to those of stainless steel and martensitic nickel-titanium (Proffit et al., 2012). The stiffness of beta-titanium is double that of nickel-titanium but still less than the stiffness of stainless steel. Beta-titanium also has greater springback than stainless steel, allowing it to be deflected nearly twice as much as stainless steel without permanent deformation. Beta-titanium delivers roughly half the amount of force when compared to stainless steel. The lower force delivery exerted by beta-titanium alloys provides the opportunity to more fully engage the bracket slot without applying more force, for example: a 0.018 x 0.025 inch beta-titanium wire delivers nearly the same force as does a 0.014 x 0.020 inch stainless steel wire (Kapila & Sachdeva, 1989). Though orthodontic treatment is not associated with an increase in nickel hypersensitivity, unless patients have a history of nickel exposure from cutaneous piercings (Kolokitha et al., 2008), beta-titanium is a great nickel-free archwire option for patients with severe pre-existing nickel allergies. The major disadvantages of beta-titanium are that it has a higher coefficient of friction than any other orthodontic alloy (Kusy, 1997) and it may be susceptible to fracture during clinical manipulation (Verstrynge et al., 2006). The combination of formability, strength and springiness allow beta-titanium to be a great intermediate and finishing archwire (Proffit et al., 2012).

Evolution of Esthetic Appliances

Prior to 1980, the only option for an orthodontist desiring to use fixed appliances was to cement a metal band on every tooth. The result was an appliance that was extremely visible and unesthetic. The development of adhesives capable of providing a good mechanical lock to the enamel surface resulted in a shift from banded to bonded appliances. Rather than fitting a band on each tooth, clinicians were now able to bond orthodontic attachments directly to the enamel surface. This development not only eased the burden of banding each tooth, but it also eliminated a significant amount of unsightly metal from the fixed orthodontic appliance (Proffit et al., 2012). The movement towards a more esthetic orthodontic appliance has been important as more adults have been seeking out orthodontic treatment (Imai et al., 1999). The demand for an increase in appliance esthetics has led to a number of esthetic treatment options, including clear plastic aligners, lingually bonded appliances and the more commonly utilized clear or translucent labial brackets. Unfortunately, the majority of esthetic labial brackets continue to be used in concert with the highly efficient, yet unesthetic, alloy archwires (Burstone et al., 2011). The next step to increase the esthetics of fixed orthodontic appliances is to use an esthetic archwire (Huang et al., 2003) in concert with clear esthetic brackets (see Figure 1).

Figure 1. Comparison of esthetics when using clear polymer brackets with an alloy archwire (maxillary arch) versus a fiber-reinforced composite archwire (mandibular arch).

Tooth colored plastic coatings, such as Teflon, placed over traditional alloy archwires has been one development aimed at increasing orthodontic archwire esthetics.

This plastic coating offers a low coefficient of friction, but unfortunately bending of the wire can be limited and the coatings can peel off or disappear within as little as three weeks due to the hostile mechanicochemical environment of the oral cavity (Burstone et al., 2011; Kusy, 2002). Another esthetic option is to fabricate transparent composite archwires (Imai et al., 1999).

Esthetic Composite Archwires

The two types of transparent polymeric composite archwires that have recently been developed are self-reinforced and fiber-reinforced. Self-reinforced composite archwires, based on a polyphenylene polymer, are fiber free and exhibit high springback, ductility, yield strength and modulus of elasticity (Goldberg et al., 2011). Translucency and good formability are additional benefits that indicate polyphenylene polymers may be an efficient and esthetic option for an orthodontic archwire material (Burstone et al., 2011).

Fiber-reinforcement of composites has been used in a variety of dental applications, such as: provisional bridges, retainers, space maintainers, orthodontic wires, endodontic posts and cores, fixed partial dentures, periodontal and orthodontic splints, as well as trauma stabilization (Goldberg & Burstone, 1992; Jancar et al., 1994; Valiathan & Dhar, 2006). Adding glass fibers to reinforce a polymer leads to increased strength and rigidity (Burstone et al., 2011). The fibers used for reinforcement may be short fibers or continuous filaments. Short fibers, usually less than 1/8 inch, are arranged parallel to the long axis of the wire and result in a wire with low stiffness. When continuous fibers are incorporated, they are aligned parallel to each other along the long axis of the wire.

Wires reinforced with continuous filaments have a large range of springback and elastic recovery. The volume percentage of fiber within the polymer wire is highly variable, ranging anywhere from 5% to 80%. As the percentage of fiber increases, so does the stiffness and yield strength. A large benefit of fiber-reinforced composite archwires is that they can be manufactured to be anisotropic. The ability to alter the fiber orientation and percentage within the polymer makes it possible to tailor wires with different properties in torsion and flexure (Valiathan & Dhar, 2006.) This enables wires to be manufactured with the same cross-sectional dimensions and yet have different stiffness values. Consequently, it is possible to achieve uniform archwire engagement into the bracket slot all throughout treatment (Valiathan & Dhar, 2006; Fallis & Kusy, 2000; Zufall & Kusy, 2000).

Manufacturing Method

Most fiber-reinforced composite archwires discussed in the literature are manufactured through a process called pultrusion (Huang et al., 2003), which was developed in 1950 by W. B. Goldsworthy (Kennedy & Kusy, 1995). The pultrusion process involves pulling fiber bundles through an extruder in which they are wetted with monomer resin. The wetted fibers then move to a die where they are formed into round or rectangular cross-sectional morphology while the monomer is cured with heat and pressure. If the monomer is only partially cured initially, the longitudinal morphology may be further shaped through a process called beta-staging (Valiathan & Dhar, 2006). Combining the pultrusion process with beta-staging provides the ability to control the longitudinal shape as well as the cross-sectional profile of the resulting fiber-reinforced composite (Kusy & Kennedy, 1999).

The fiber-reinforced composite archwires used in this research project were fabricated by pultrusion using a vertically oriented, shrinkable and flexible die. A composite of fibers and resin is pulled into the die, which is compressed as the die shrinks to the determined cross-sectional size and shape. A flexible die is used to allow the composite to be bent lengthwise into the desired longitudinal shape prior to curing (Gopal et al., 2005). The resulting wires are then packaged and marketed by BioMers Products.

Viscoelastic Properties of Fiber-reinforced Composite Archwires

Although esthetics are desired by patients and orthodontists alike, proper and efficient function of the appliance is mandatory (Kusy, 1997). When a constant deflection is applied to an alloy archwire, the amount of force delivered will remain constant. Polymer based composite archwires typically exhibit time-dependent stress-strain behavior which may lead to decreased force delivery over time (Figure 2). This decrease in force delivery, known as stress relaxation, is due to relaxation of the molecular confirmations toward equilibrium, despite the constant deflection (Goldberg et al., 2011). Clinically, a decrease in force delivery over time would lead to inefficient tooth movement if the force levels decrease below the minimum threshold for tooth movement (Valiathan & Dhar, 2006).

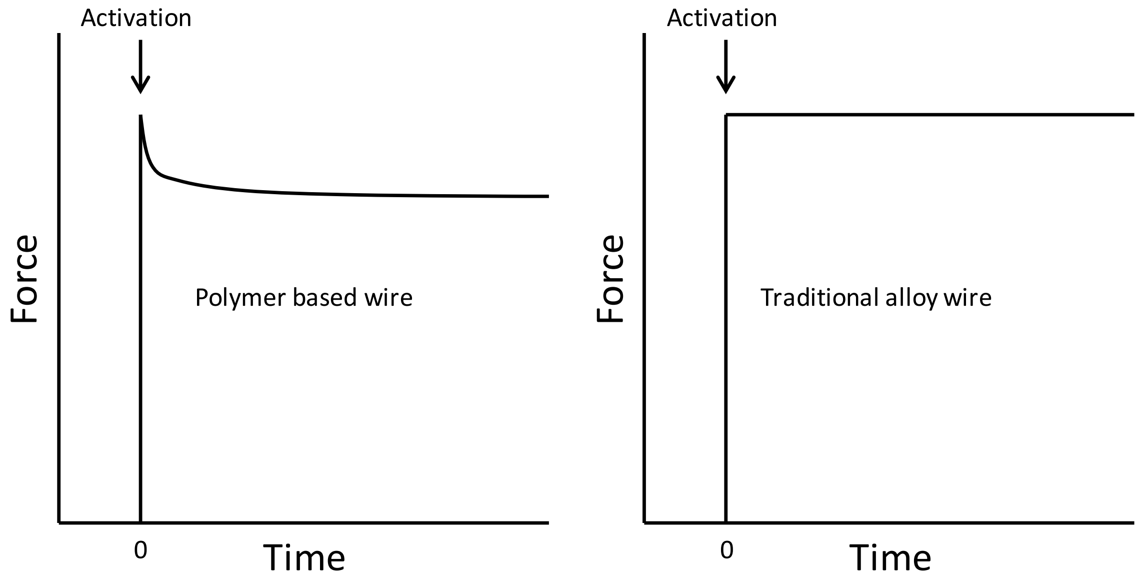

Figure 2. Comparison of changes in force levels with a constant activation of a polymer based wire experiencing stress relaxation and a traditional alloy wire where the force remains constant (adapted from Goldberg et al, 2011).

Due to the potential for stress relaxation to occur in fiber-reinforced composite archwires, it is important to investigate their mechanical abilities and verify that they can sustain sufficient force levels suitable for efficient tooth movement.

CHAPTER 3
MATERIALS AND METHODS

In this study, 0.018” (Align A) fiber-reinforced composite archwires from BioMers Products and 0.016” Nitinol Classic archwires from 3M Unitek were used (Figure 3). Larger dimensions of fiber-reinforced composite archwires are available from BioMers Products, however, previous research has shown that the smallest wire (Align A) is more flexible and less likely to experience cracks or crazing during 3-point bending tests (Chang, 2012). Additionally, the smaller 0.016” Nitinol Classic wires were used because it has bending values closer to Align A compared to 0.018” Nitinol Classic (Ballard et al., 2012; Chang, 2012).

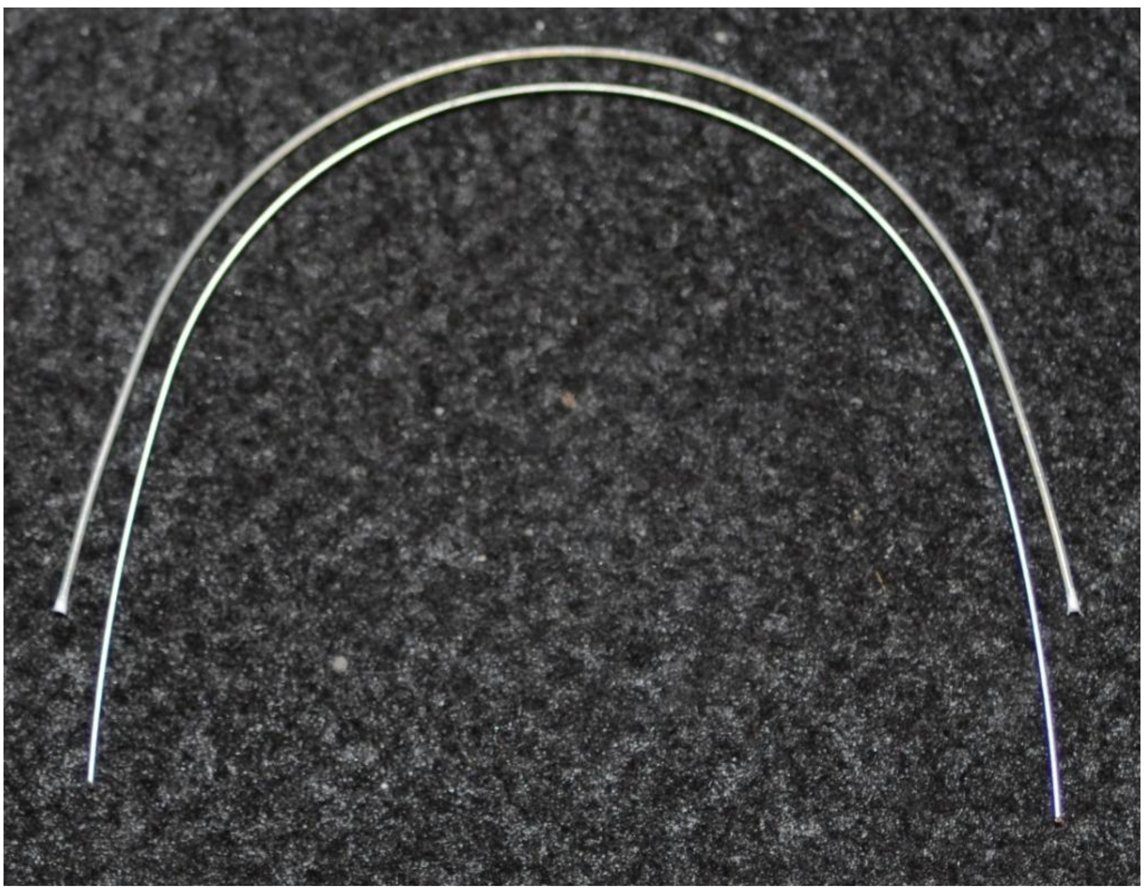

Figure 3. Photo of a 0.018” Align A fiber-reinforced composite archwire (top) and a 0.016” Nitinol Classic archwire (bottom).

This study examined the force decay (or stress relaxation) properties of BioMers esthetic fiber-reinforced composite orthodontic archwires with Nitinol Classic archwires as a comparison group. Force decay was determined utilizing a 3-point bend test to measure the amount of force necessary to deflect a specimen. Fifteen archwires of each brand were used. For each archwire, two 25 mm segments were sectioned from the distal ends of each archwire and allocated to one of two groups (1 or 2 mm groups; n=15/group). Each segment was tracked during all procedures. Segments were projected onto a screen along with a 2-dimensional Cartesian grid comprised of 0.05 x 0.05 inch squares to measure the curvature of the segments. This was performed to determine the amount of curvature and/or deformation, if any, before initial testing, after the first 3-point bend test, and after deflection for 30 days (mentioned below) to assure consistent bending configurations during testing. Curvature, the inverse of radius, was measured by fitting a circle of the same arc length as the segments to the grid. Due to the impracticality of measuring force decay of a single archwire for 30 days, the following protocol was used: each segment was tested in 3-point bending (14 mm span length; 2.0 mm/min crosshead speed; 37^{o}C in air; Figures 4-5) using a universal testing machine (Instron, Norwood, MA) to a maximum deflection of 3.1 mm (ADA Specification #32); next, each segment was placed in a custom-made jig (Figures 6-7) designed to deflect each segment either 1 or 2 mm for 30 days in air at 37^{o}C. Upon removal from the jig at 30 days, each segment was once again tested in 3-point bending to examine consistency of the bending profile. The slope (g/mm) of the linear portion of the force versus deflection curve and force (g) values at 1.0, 2.0 and 3.0 mm during both activation and deactivation comprise the data harvested from each test. Paired t-tests were used to statistically compare pre- and post- deflection forces (α =0.05). Additionally, a control group consisting of wires not subject to the 30 day constant deflection was also tested to ensure that the initial 3-point bend test did not alter the material and impact the results from the second 3-point bend test after 30 days.

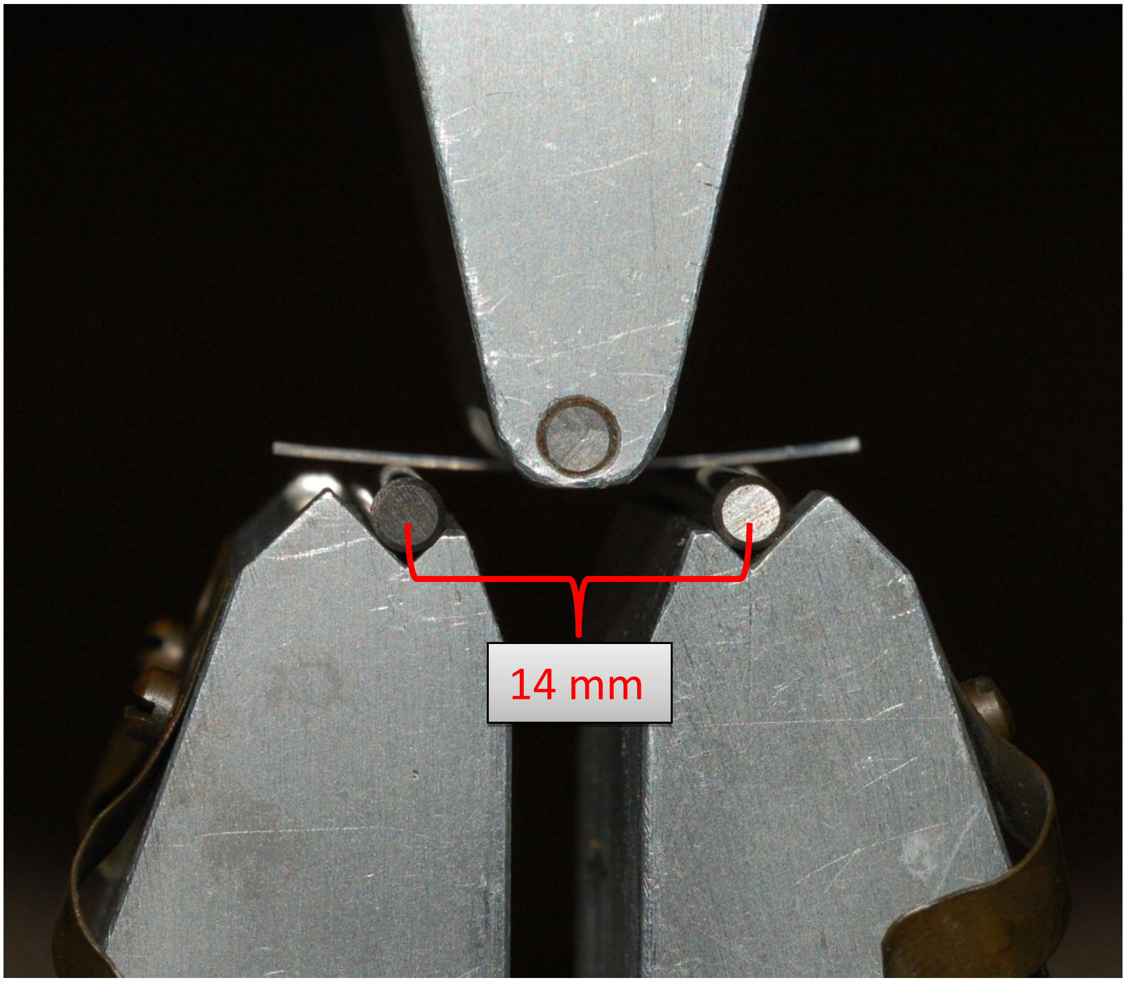

Figure 4. Testing set-up for 3-point bending. A 14 mm span length between lower supports was used with the upper beam centered at 7 mm.

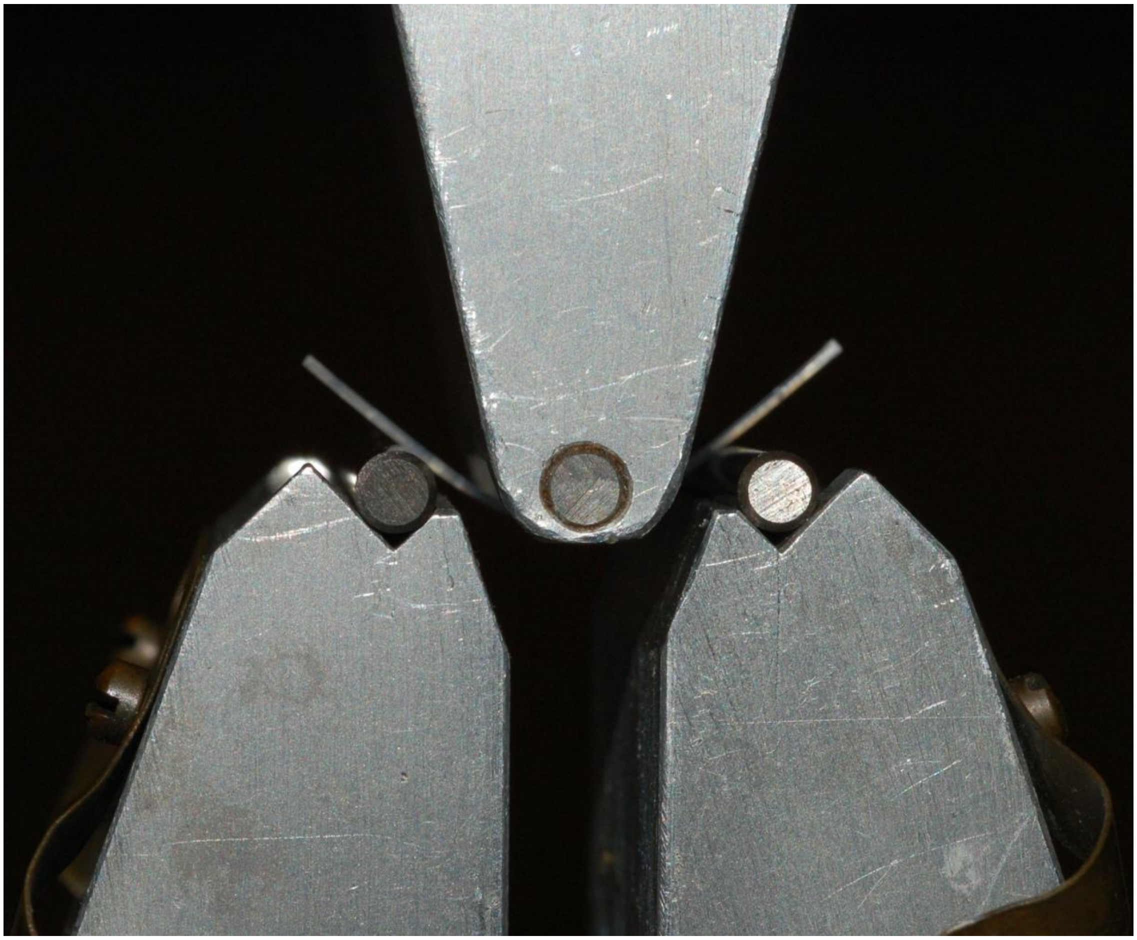

Figure 5. Three-point bending test in progress.

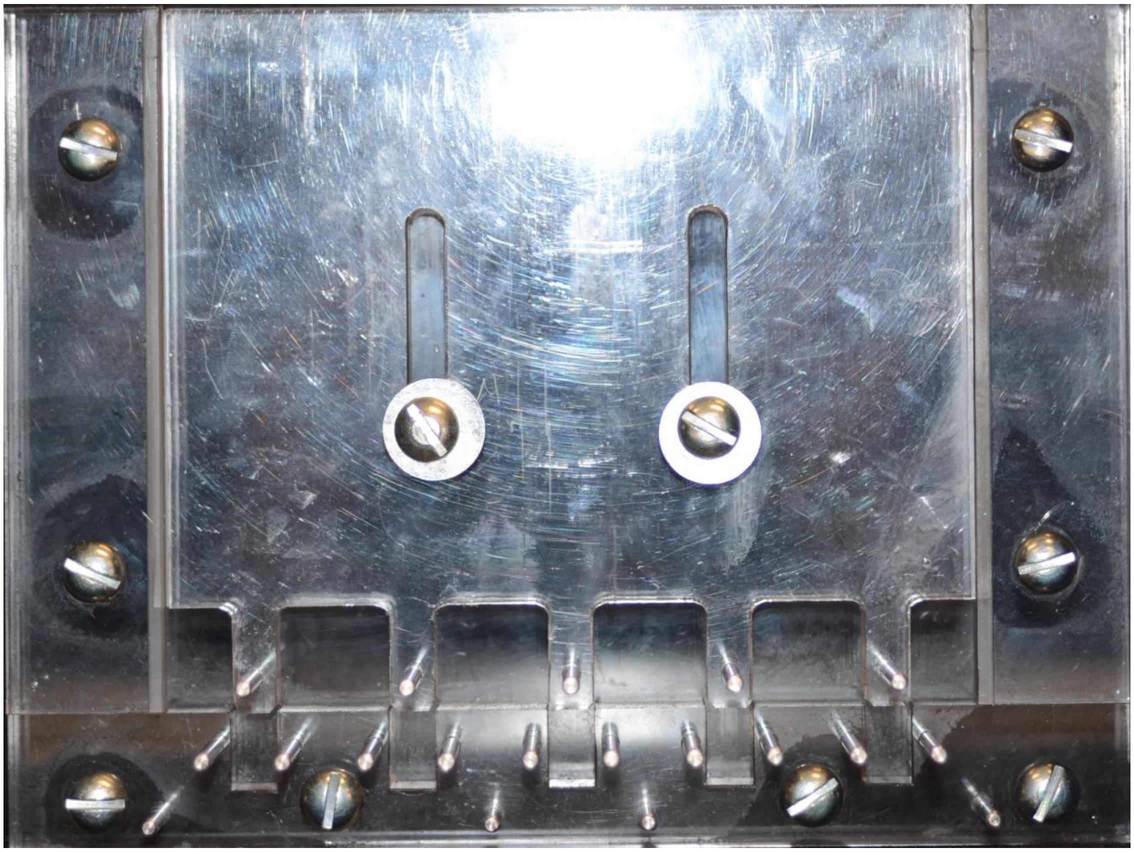

Figure 6. Custom made deflection jig. A 14 mm span length between lower supports was used with the upper beam centered at 7 mm.

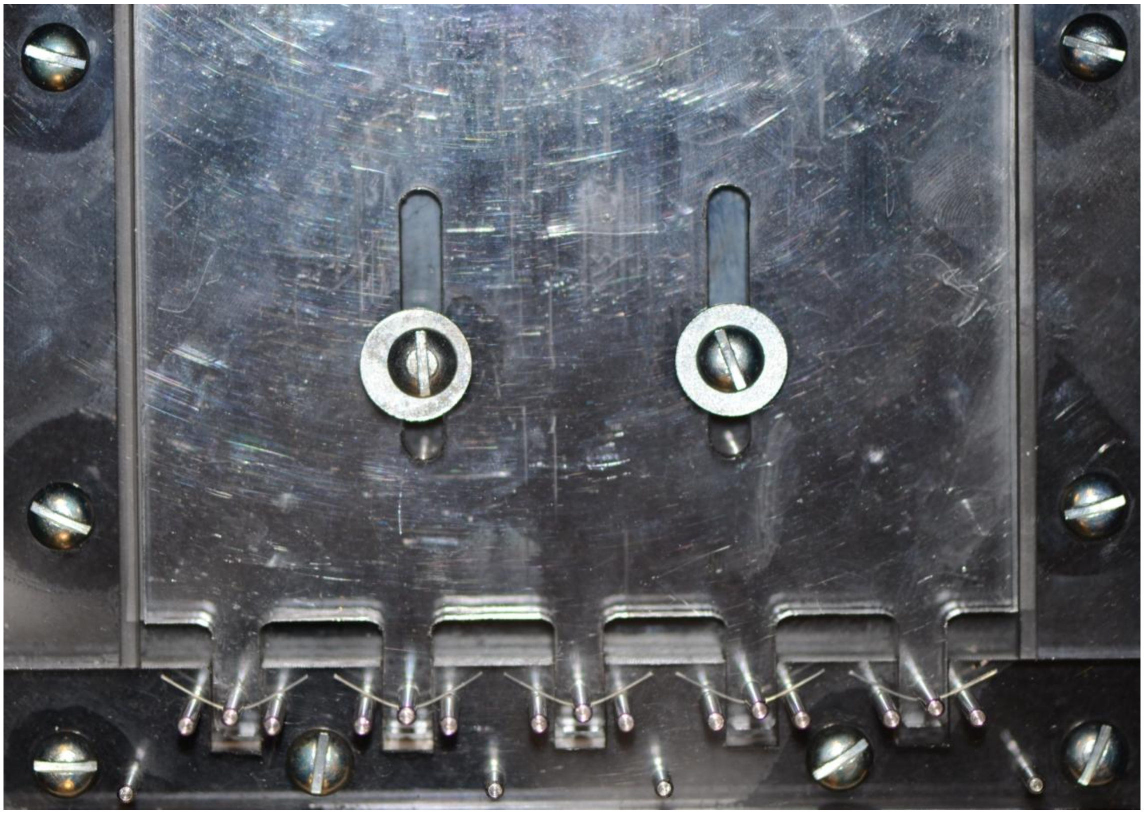

Figure 7. Testing set-up with fiber-reinforced composite wires in deflection jig.

CHAPTER 4
RESULTS

The curvatures of the fiber-reinforced composite and NiTi wire segments used in this testing were determined to be 0.01 mm^{-1} or less, which was the approximate lower sensitivity limit using the 2-dimensional Cartesian grid described above. Nevertheless, the segments did not increase in curvature after initial 3-point bending or after 30 days of deflection.

The observed bending profiles of fiber-reinforced composite archwires show similar force-deflection curves as those of nickel-titanium archwires, only with slightly lower forces observed in the fiber-reinforced composite groups (Figure 8).

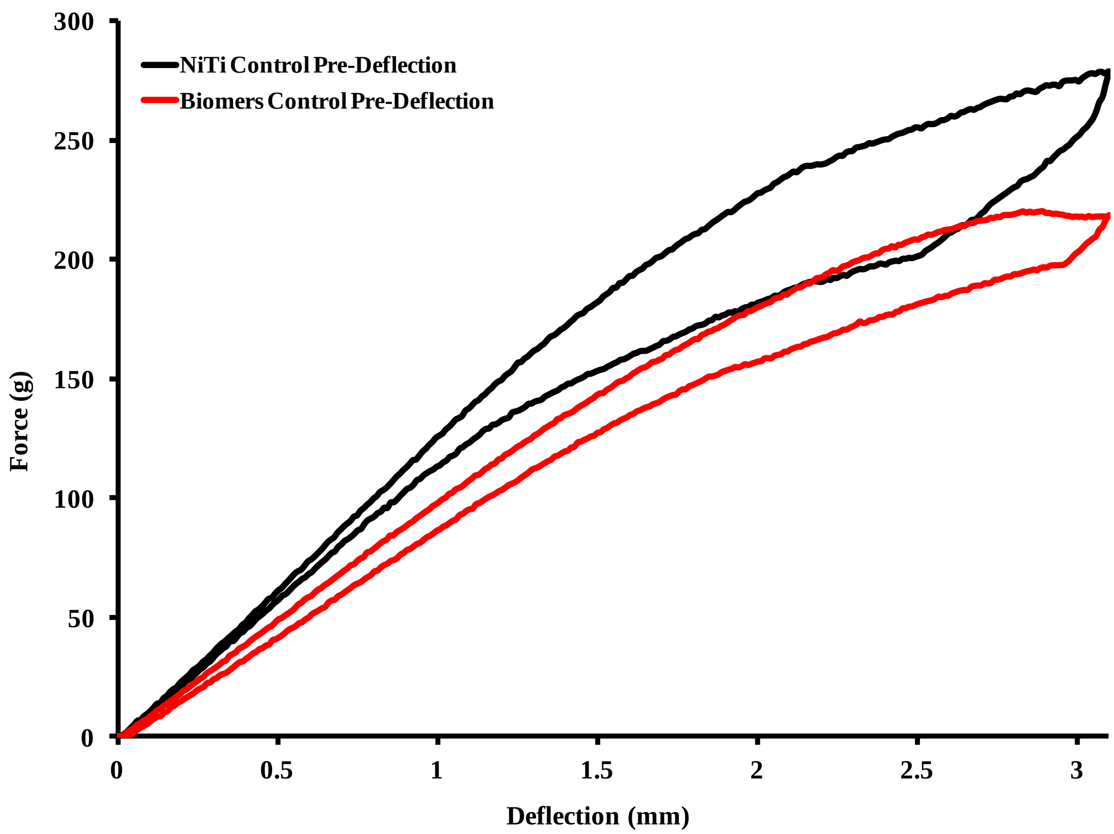

Figure 8. Comparison of typical force-deflection curves of nickel-titanium archwires and fiber-reinforced composite archwires.

The force-deflection curves obtained for each of the NiTi test groups exhibited similar activation and deactivation curves for the pre-deflection and post-deflection bending profiles (Figures 9-11). Actual activation and deactivation force values can be found in Table 1 and Table 2, respectively.

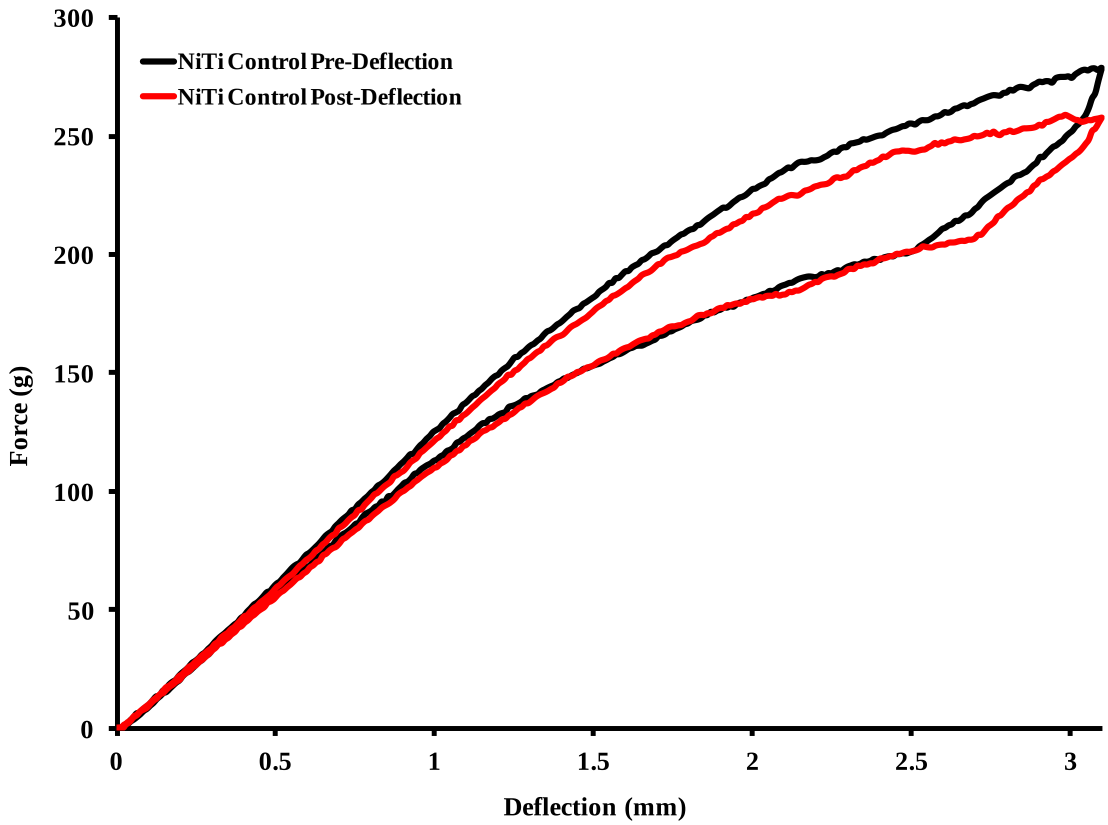

Figure 9. Comparison of typical force-deflection curves for the NiTi control group.

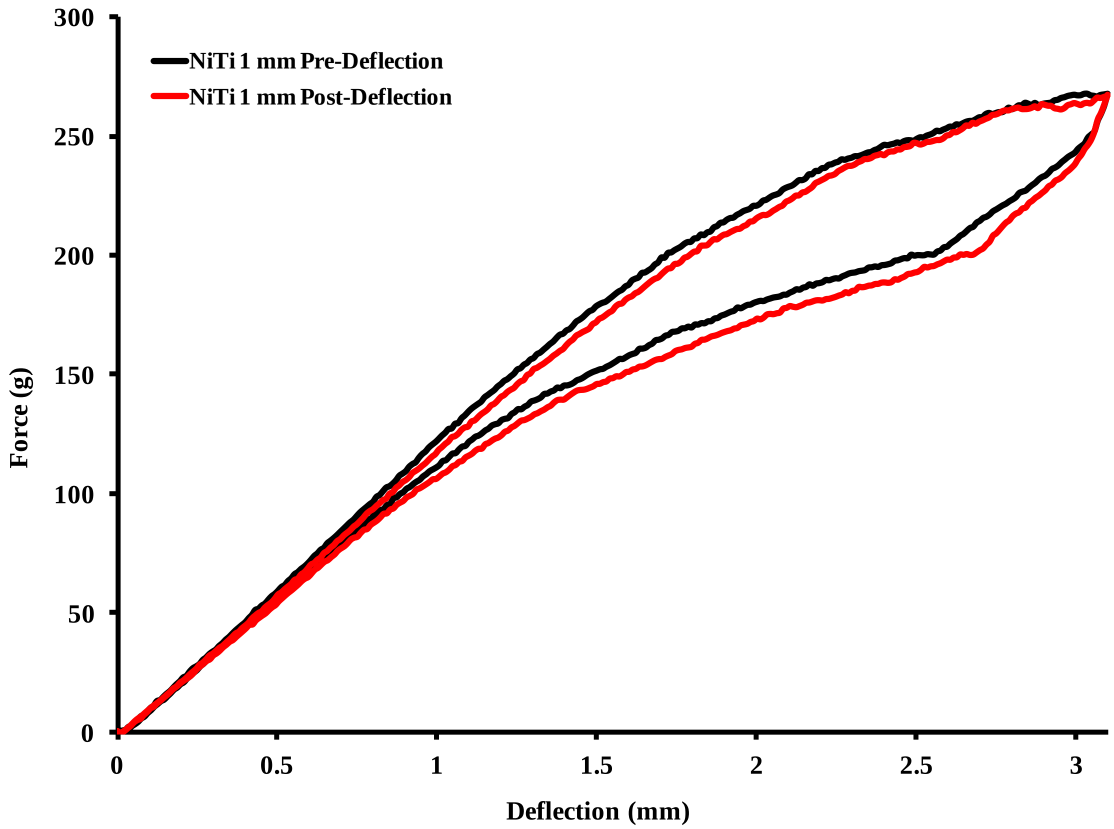

Figure 10. Comparison of typical force-deflection curves for the NiTi 1 mm deflection group.

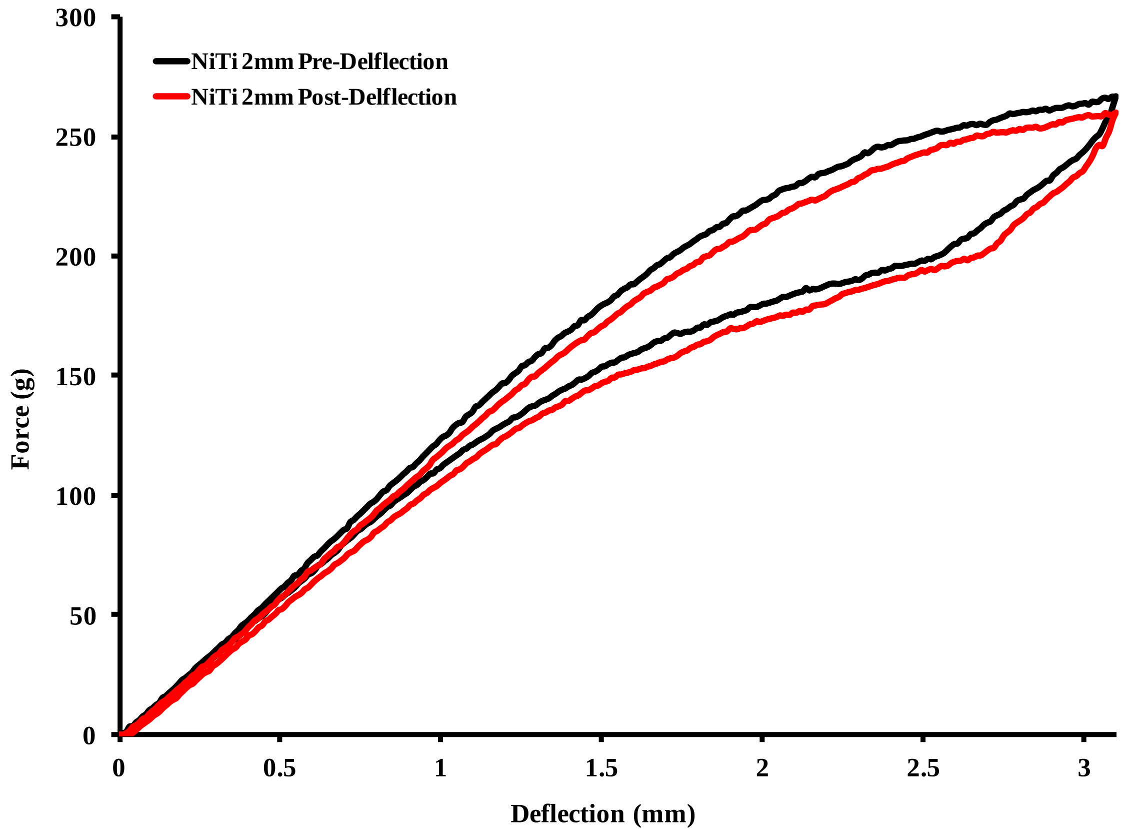

Figure 11. Comparison of typical force-deflection curves for the NiTi 2 mm deflection group.

Table 1. Bending values during activation.

ARCHWIRE	ACTIVATION					
	Stiffness (g/mm)	Modulus (GPa)	Force at 1 mm (g)	Force at 2 mm (g)	Force at 3 mm (g)	# with cracks (after bend test for pre-deflection groups, after deflection for post-deflection groups)
NiTi Control: pre-deflection	125.6 ± 2.0	56.0 ± 0.9	123.3 ± 1.4	224.2 ± 2.0	270.3 ± 5.7	0
NiTi Control: post-deflection	121.4 ± 2.3*	54.1 ± 1.0*	119.6 ± 2.7*	215.2 ± 2.6*	256.6 ± 4.3*	0
NiTi 1mm group: pre-deflection	125.5 ± 1.7	56.0 ± 0.8	123.1 ± 1.8	223.4 ± 2.5	267.2 ± 3.3	0
NiTi 1mm group: post-deflection	120.4 ± 1.8*	53.7 ± 0.8*	118.8 ± 1.4*	215.7 ± 3.0*	262.0 ± 5.7*	0
NiTi 2mm group: pre-deflection	125.7 ± 1.4	56.1 ± 0.6	123.9 ± 1.1	223.9 ± 1.5	268.3 ± 3.8	0
NiTi 2mm group: post-deflection	119.5 ± 1.4*	53.3 ± 0.6*	117.6 ± 1.0*	213.0 ± 2.2*	255.9 ± 4.7*	0
BioMers Control: pre-deflection	101.1 ± 9.1	27.2 ± 2.4	98.8 ± 9.6	181.5 ± 17.1	219.5 ± 18.6	1
BioMers Control: post-deflection	99.1 ± 8.6	26.6 ± 2.3	96.1 ± 7.6	176.7 ± 12.9	217.1 ± 14.8	1
BioMers 1mm group: pre-deflection	97.4 ± 18.8	26.2 ± 5.1	94.2 ± 18.3	175.9 ± 34.7	205.1 ± 52.3	2
BioMers 1mm group: post-deflection	87.5 ± 23.4	23.5 ± 6.3	85.7 ± 22.7	158.3 ± 41.5	193.6 ± 48.6*	2
BioMers 2mm group: pre-deflection	99.9 ± 15.2	26.8 ± 4.1	97.7 ± 14.6	177.4 ± 27.8	217.4 ± 31.8	2
BioMers 2mm group: post-deflection	47.9 ± 39.0*	12.9 ± 10.5*	46.8 ± 37.9*	85.8 ± 68.5*	105.9 ± 83.0*	12

Within each parameter, * denote significant differences ($p<0.05$) exist between pre- and post-deflection wires.

Table 2. Bending values during deactivation.

ARCHWIRE	DEACTIVATION					
	Stiffness (g/mm)	Modulus (GPa)	Force at 3 mm (g)	Force at 2 mm (g)	Force at 1 mm (g)	Elastic Recovery (%)
NiTi Control: pre-deflection	118.4 ± 1.4	52.8 ± 0.6	247.7 ± 2.8	179.0 ± 2.6	111.9 ± 1.4	99.1 ± 0.6
NiTi Control: post-deflection	113.7 ± 1.7*	50.7 ± 0.8*	238.9 ± 2.4*	177.2 ± 2.1*	108.7 ± 2.1*	99.4 ± 0.4
NiTi 1mm group: pre-deflection	118.0 ± 2.2	52.6 ± 1.0	249.4 ± 8.7	180.7 ± 3.5	114.4 ± 8.2	99.5 ± 0.5
NiTi 1mm group: post-deflection	113.3 ± 1.7*	50.5 ± 0.7*	240.8 ± 2.9*	175.3 ± 2.4*	108.1 ± 2.0*	99.3 ± 0.6
NiTi 2mm group: pre-deflection	119.6 ± 1.9	53.3 ± 0.8	247.7 ± 2.9	180.4 ± 2.4	112.7 ± 1.1	99.2 ± 0.4
NiTi 2mm group: post-deflection	112.2 ± 1.1*	50.0 ± 0.5*	237.9 ± 3.3*	174.3 ± 2.4*	106.2 ± 1.8*	98.8 ± 0.6
BioMers Control: pre-deflection	89.6 ± 6.3	24.1 ± 1.7	201.2 ± 12.7	156.6 ± 10.3	85.9 ± 6.4	99.0 ± .07
BioMers Control: post-deflection	89.1 ± 7.2	23.9 ± 1.9	200.0 ±13.1*	155.6 ± 11.0	85.3 ± 7.0	99.1 ± 0.7
BioMers 1mm group: pre-deflection	80.3 ± 24.4	21.6 ± 6.6	187.3 ± 49.2	139.7 ± 39.9	76.0 ± 23.2	98.5 ± 1.4
BioMers 1mm group: post-deflection	77.4 ± 21.3	20.8 ± 5.7	177.7 ± 45.2*	135.5 ± 37.4	74.2 ± 21.4	98.8 ± 1.1
BioMers 2mm group: pre-deflection	82.7 ± 24.5	22.2 ± 6.6	196.2 ± 35.8	143.8 ± 41.3	79.0 ± 23.8	98.6 ± 2.3
BioMers 2mm group: post-deflection	37.4 ± 37.7*	10.1 ± 10.1*	94.0 ± 78.8*	65.6 ± 66.0*	35.6 ± 37.4*	99.1 ±1.9

Within each parameter, * denote significant differences ($p<0.05$) exist between pre- and post-deflection wires.

Statistically significant ($p<0.05$) differences in the pre-deflection and post-deflection stiffness values, during activation and deactivation, were evident in each of the NiTi test groups. The activation and deactivation force levels measured in the NiTi test groups were very consistent, resulting in small standard deviations (Tables 1-2). Though the small decrease in post-deflection NiTi force levels is statistically significant, this small decrease in force is not clinically significant (Figures 12-13).

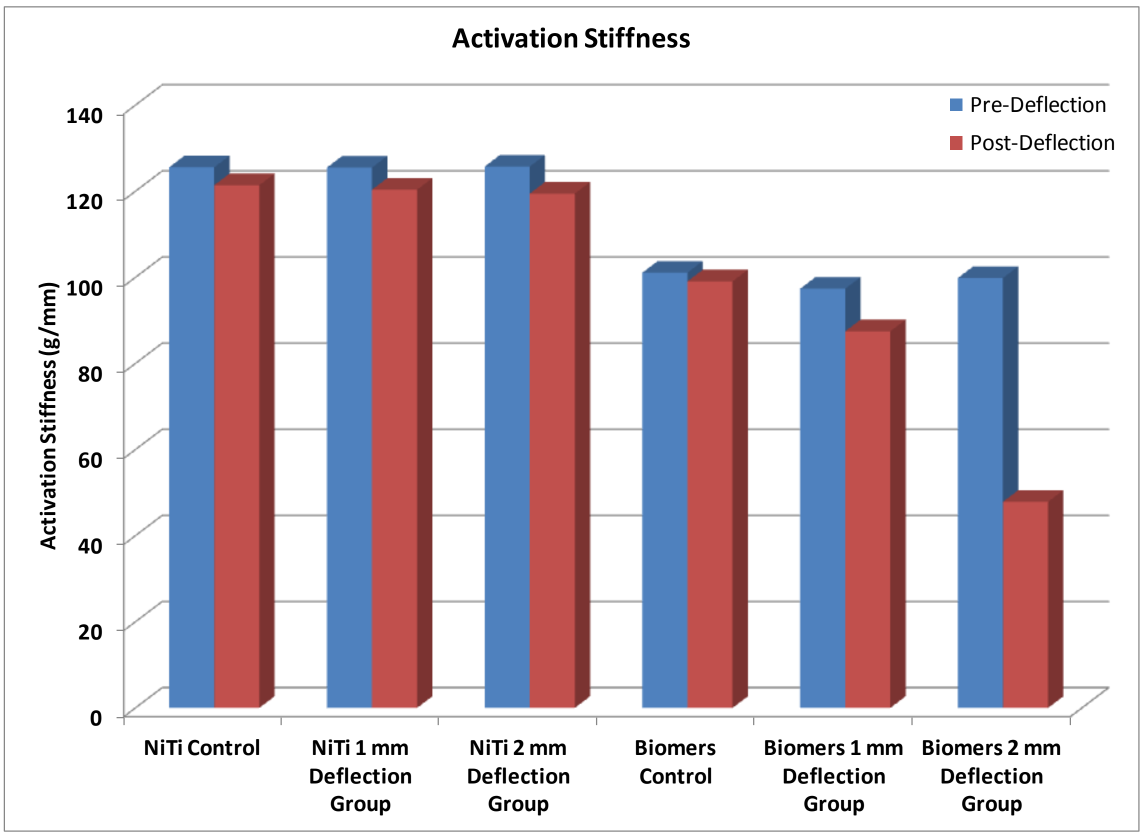

Figure 12. Activation stiffness (g/mm) for each test group.

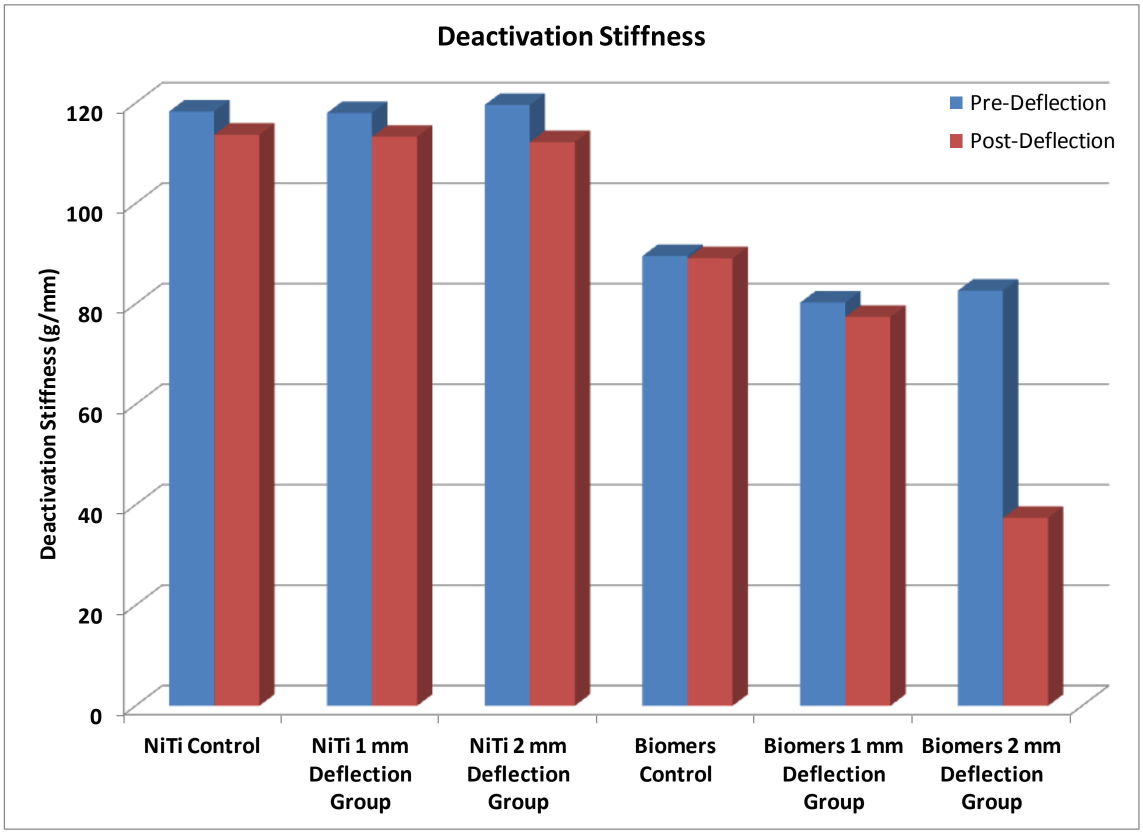

Figure 13. Deactivation stiffness (g/mm) for each test group.

Similar activation and deactivation curves, for the pre-deflection and post-deflection bending profiles, were found in the BioMers control group as well as the BioMers 1 mm deflection group (Figures 14-15). The differences in the pre-deflection and post-deflection activation and deactivation values were not statistically significant (($p>0.05$); Tables 1-2).

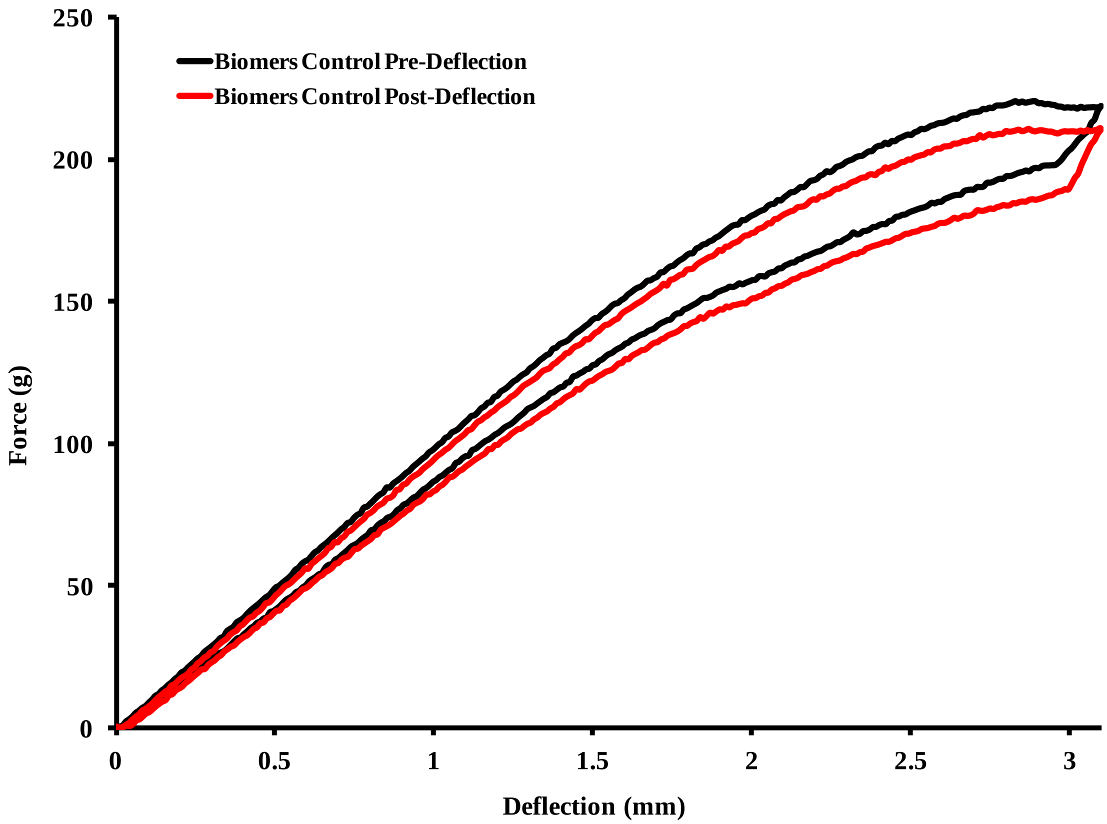

Figure 14. Comparison of typical force-deflection curves for the fiber-reinforced composite control group.

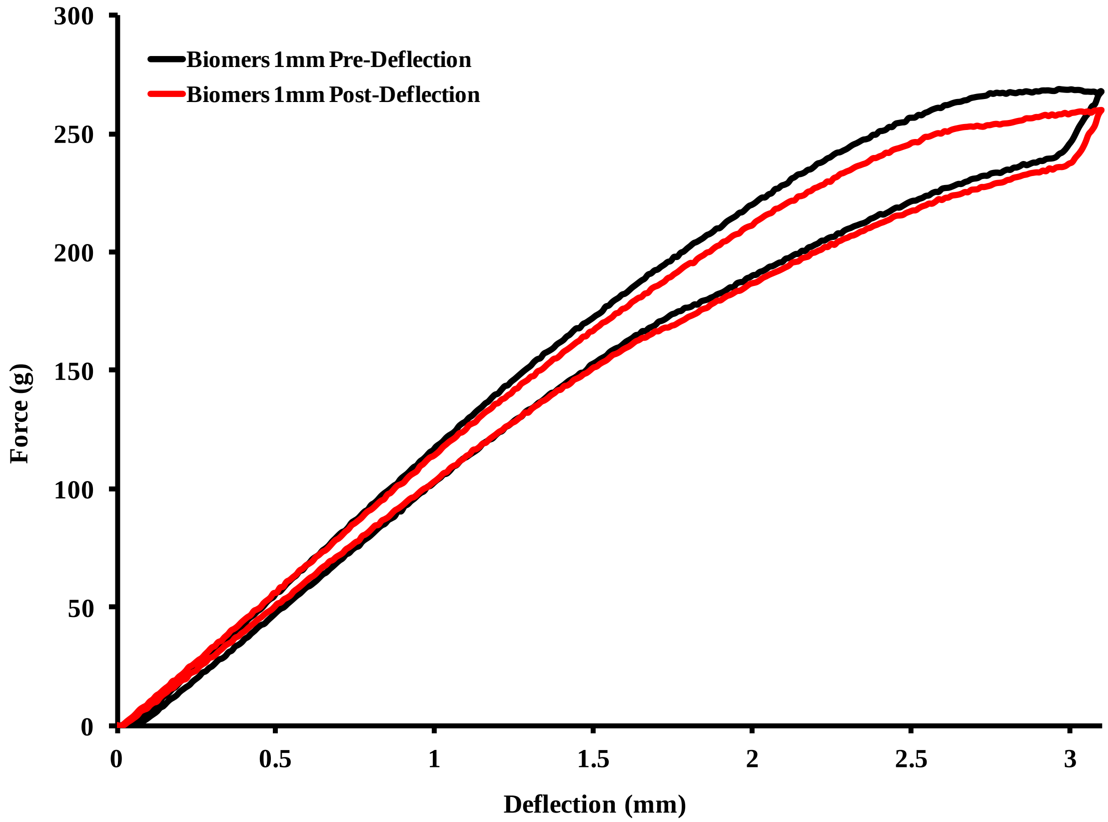

Figure 15. Comparison of typical force-deflection curves for the fiber-reinforced composite 1 mm deflection group.

Statistically significant ($p<0.05$) differences in the pre-deflection and post-deflection stiffness, during activation and deactivation, were evident in the BioMers 2 mm deflection group. The BioMers 2 mm deflection group failed to deliver consistent forces as 80% of the wires experienced varying degrees of crazing during the 30 day deflection period (Figure 16). The post-deflection force levels measured in the BioMers 2 mm group were highly variable. The activation and deactivation force levels for the few wires that did not experience crazing were close to pre-deflection values (Figure 17), whereas the crazed wires exhibited large decreases in activation and deactivation force levels (Figures 18-19).

As mentioned above, the curvature of the tested wire segments returned to its as-received shape when projected along with a calibrated grid, indicating that they were not deformed by being stored deflected for 30 days.

Figure 16. Comparison of non-crazed (top) and crazed (bottom) fiber-reinforced composite archwire.

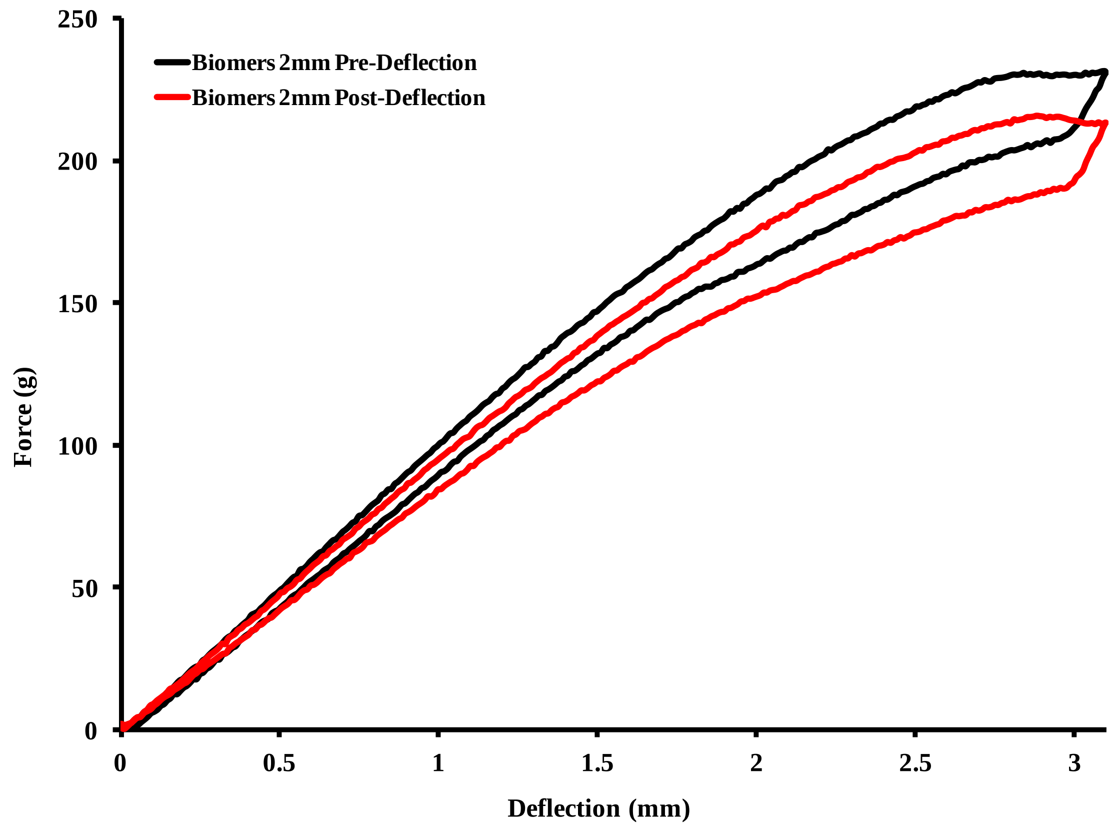

Figure 17. Comparison of force-deflection curves for a non-crazed fiber-reinforced composite wire in the 2 mm deflection group.

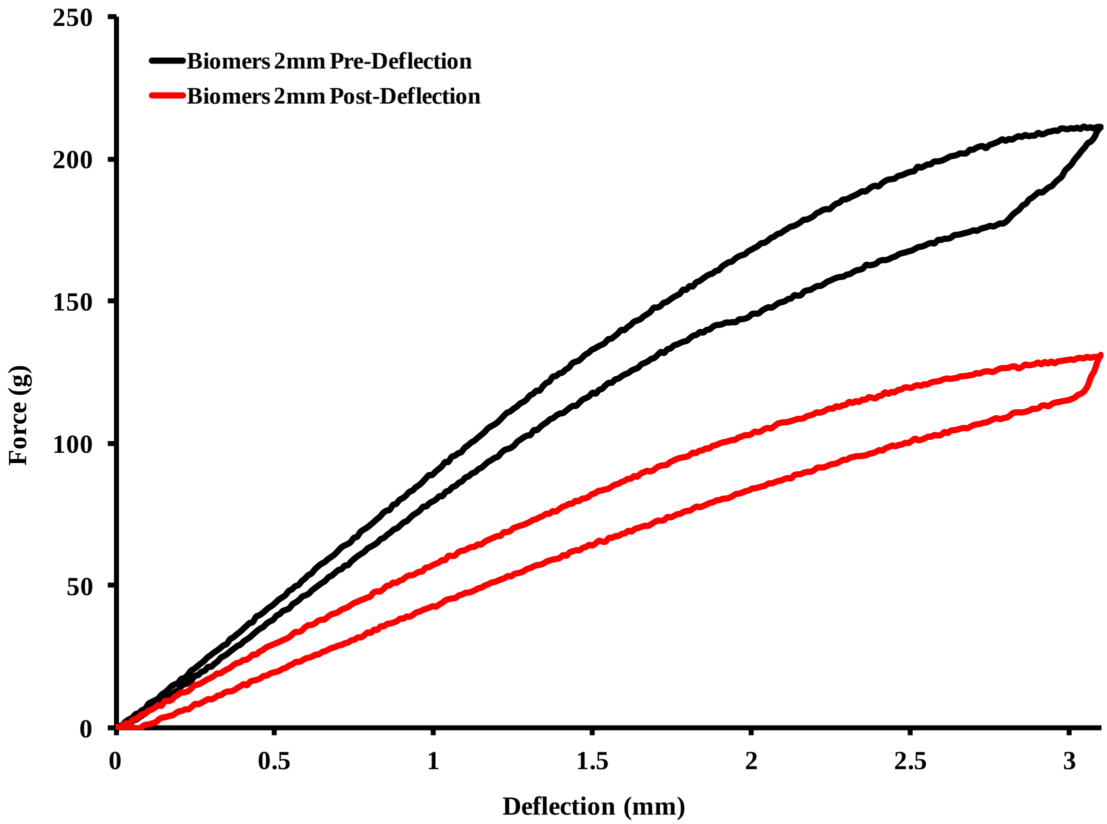

Figure 18. Comparison of force-deflection curves for a crazed fiber-reinforced composite wire in the 2 mm deflection group.

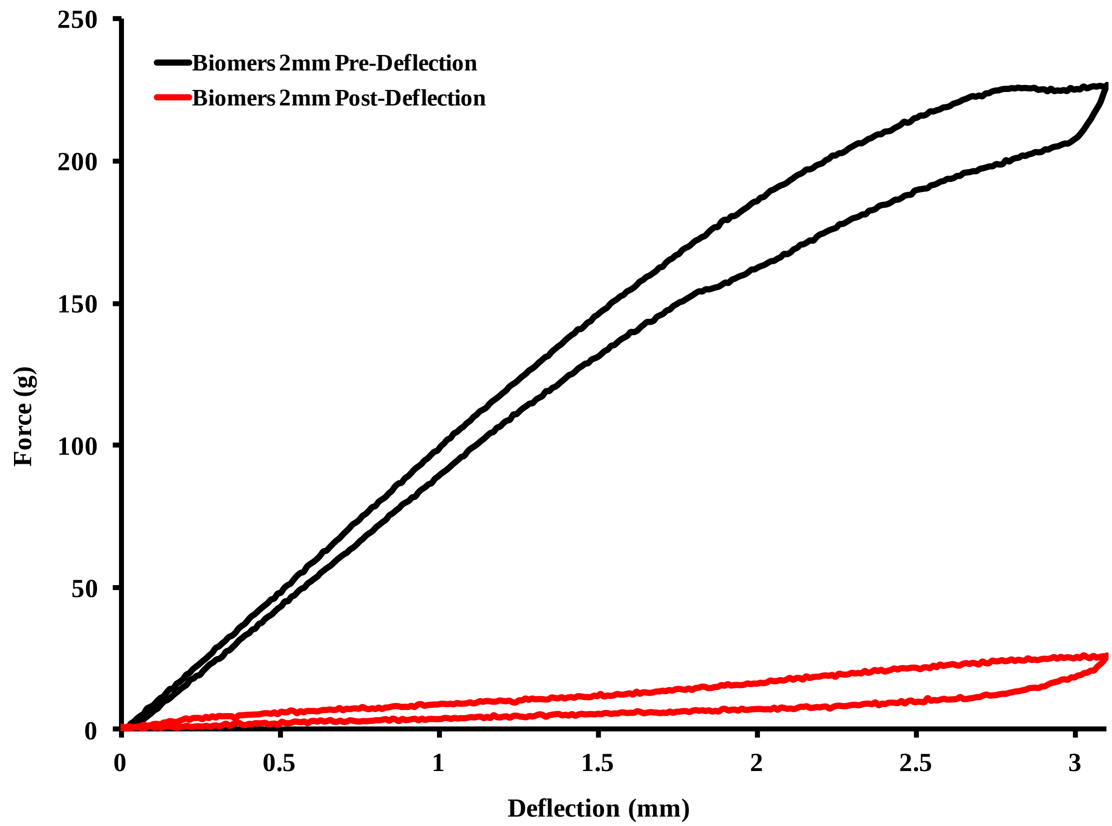

Figure 19. Comparison of force-deflection curves for a crazed fiber-reinforced composite wire in the 2 mm deflection group exhibiting very low force levels.

CHAPTER 5
DISCUSSION

Fiber-reinforced composite materials have a variety of different applications within the field of dentistry (Cacciafesta et al., 2008; Jancar et al., 1994). The use of a fiber-reinforced composite archwire in concert with an esthetic polymer or ceramic bracket would serve to increase the esthetics of the fixed orthodontic appliance. Fiber-reinforced composite archwires can also provide practitioners with a nickel-free treatment option when presented with patients exhibiting severe nickel allergies (Valiathan & Dhar, 2006). Since fiber-reinforced composite archwires can be manufactured to be anisotropic, it is possible to alter the stiffness values of an archwire without changing its cross-sectional dimensions. This ability makes it possible to more fully engage the bracket slot early in treatment and subsequently maintain the desired engagement throughout treatment (Valiathan & Dhar, 2006; Zufall & Kusy, 2000). In order for practitioners to be able to take full advantage of these benefits, a fiber-reinforced composite archwire must exhibit clinically effective mechanical properties.

This study found the bending properties of fiber-reinforced composite archwires to be similar to those of nickel-titanium archwires. When comparing wires with similar cross-sections, the fiber-reinforced composite archwires deliver lower force levels than nickel-titanium archwires. This can be observed when comparing the force-deflection curves of each respective material (Figure 8). These findings are in harmony with a recent study that found while fiber-reinforced composite archwires are less stiff and deliver less force than nickel-titanium archwires of the same dimension, they have bending properties similar to nickel-titanium and force levels within the same range (Ballard et al., 2012).

Nickel-titanium archwires are time tested and have a record of great clinical efficacy due to their high springback, flexibility and resistance to plastic deformation as well as the ability to maintain a continuous light force over a long range of time, regardless of the amount of deflection (Kapila & Sachdeva, 1989). For fiber-reinforced composite archwires to be considered as a viable treatment alternative for nickel-titanium archwires they must not experience large amounts of stress relaxation and they must be able to undergo large deflections without permanently deforming or crazing. The results from the BioMers 1 mm deflection group showed that fiber-reinforced composite archwires are able to deliver consistent force levels following a long period of deflection (Figure 15). However, the results from the BioMers 2 mm deflection group demonstrate that fiber-reinforced composite archwires are unable to predictably resist crazing when being deflected 2 mm over a long period of time, resulting in delivery of inconsistent force levels. Of the 15 segments tested in the BioMers 2 mm deflection group, seven experienced severe crazing during the 30 day deflection period and exhibited extremely low force levels in the post-deflection 3-point bending tests (Figure 19). Moderate force levels were observed in four of the crazed segments (Figure 18) and force levels similar to pre-deflection values were measured in one crazed segment and the three segments that did not craze during testing (Figure 17). The large variation observed within the BioMers 2 mm test group is the reason the standard deviations for this group are so high (Table 1-2). The clinical applicability of these fiber-reinforced composite archwires may be limited since only 20% of the wires in the BioMers 2mm deflection group were able to resist crazing/cracking during prolonged deflection and subsequently maintain their initial force levels.

It should be noted that the term crazing is used here to describe the structural change in the fiber-reinforced composite archwires because that term accurately describes the appearance of the wire (Figure 16), i.e. whitening of the wire, consistent with how crazing appears in polymer-based materials. Additionally, the manufacturer's literature describes the process as crazing when excessive forces cause the resin to crack. In the wires tested in this study, the exact failure mechanism was not explored. It may well be that the resin surrounding the reinforcing fibers cracking is the cause of the crazing appearance. Another possible explanation is that when fiber-reinforced composite archwires undergo long periods of deflection, the constant strain causes the interface of the fibers and polymer matrix to fail, which then transfers the load to the brittle fibers, resulting in fracture of the fibers. Failure analysis via microscopy or other techniques appears warranted to investigate the cause of the crazing and associated drop in force values.

During the initial 3-point bend test, each wire segment was deflected 3.1 mm. While only two of the wire segments from the BioMers 2 mm deflection group crazed due to the 3.1 mm deflection, twelve wire segments experienced variable amounts of crazing while being stored at a deflection of 2 mm (Table 1; Figure 16). This suggests that there is a period of time in which fiber-reinforced composite archwires are able to successfully withstand deflections of 2 mm or greater before they fail. As it was impractical to measure the force levels exerted by a deflected archwire for a period of 30 days, it is unclear when during the deflection period each of these wires crazed. If data was available regarding when each wire failed during the 30 day deflection period, it could provide insight as to how long a practitioner could leave these wires in place and expect them to provide reasonably effective force levels. Additionally, as force is transferred from the wire to the teeth, the resulting tooth movement will serve to decrease the deflection of the wire. Because of the time-dependent stress-strain behavior exhibited by polymeric wires, it is possible to recover a portion of the deformation and the force loss once the deflection is decreased (Goldberg et al., 2011). It is also possible that a reduction in the amount of deflection may result in fewer crazes/cracks and more consistent force delivery.

In this study the statistically significant ($p<0.05$) differences in each of the NiTi test groups were unexpected. As mentioned previously, the force levels in the NiTi test groups were very consistent, resulting in small standard deviations within each test group. Thus, the statistically significant difference may be attributed to the small standard deviations. Force levels necessary for tooth movement, which varies depending on the type of movement desired, are typically in the 50 gram range but can be as low as 10 grams (Proffit et al., 2012). In the NiTi test groups the average difference between pre-deflection and post-deflection stiffness (g/mm), for activation and deactivation, was less than 6 g/mm resulting in average stiffness levels of approximately 120 g/mm (Tables 1-2); thus it is evident that though the measured force levels were reduced by a statistically significant amount, the decrease in force observed in the NiTi groups was not clinically significant.

CHAPTER 6
CONCLUSION

This study demonstrated that fiber-reinforced composite archwires exhibit mechanical properties similar to those of nickel-titanium archwires when subjected to 3-point bending tests. Following 30 days of continuous 1 mm deflection, fiber-reinforced composite archwires do not exhibit significant amounts of force decay as they are able to deliver post-deflection force levels consistent with their pre-deflection force levels. However, the clinical applicability of fiber-reinforced composite archwires may be limited as the majority of the tested wires were unable to sustain deflections of 2 mm without crazing and experiencing a statistically and clinically significant decrease in force delivery.

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