The Effect of Water Storage on Bending Properties of Esthetic, Fiber-Reinforced Composite Orthodontic Wires

Ju-Han Chang
Marquette University

THE EFFECT OF WATER STORAGE ON BENDING PROPERTIES OF ESTHETIC, FIBER-REINFORCED COMPOSITE ORTHODONTIC WIRES

by

Ju-Han Chang, D.D.S.

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ABSTRACT

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Introduction: The translucent optical property of fiber-reinforced composite wires meets the esthetic demand of orthodontic patients; however, studies need to be conducted to determine if they also have the desired mechanical properties for active orthodontic treatment. The goal of this research was to study the effect of water storage on the mechanical properties of fiber-reinforced composite archwires and compare it to conventional nickel-titanium (NiTi) and stainless steel (SS), and beta-titanium (TMA) archwires.

Materials and Methods: Align A, B, C and TorQ A, B from BioMers Products, 0.014", 0.016", 0.018", 0.019" x 0.025" Nitinol Classic (3M Unitek), 0.016" SS, and 0.019 x 0.025 TMA archwires were tested in this study (n=10/type/size/condition). A 20 mm segment was cut from each end of the archwire with one then stored in distilled water at 37°C for 30 days while the other was stored dry. The segments were tested at 37±2°C using 3-point bending to a maximum deflection of 3.1 mm with force monitored during loading (activation)/unloading (deactivation). ANOVA and paired t-tests were used for statistical analysis.

Results: In terms of stiffness and force delivery during activation, in general 0.019" x 0.025" TMA > TorQ B > TorQ A > 0.019" x 0.025" NiTi > 0.016" SS > Align C > 0.018" NiTi > Align B > 0.016" NiTi > Align A > 0.014" NiTi. Water exposure was detrimental to the larger translucent wires (Align B and C) as they were more likely to crack/craze during bending, resulting in decreased amounts of force applied at a given deflection. All TorQ A and B wire segments cracked during the test; the stored in water groups had significantly greater decrease in force level delivery. Align A and the alloy wires were not significantly affected by water storage. Overall, the alloy wires possessed vastly more consistent force values compared to the composite wires.

Conclusions: Although the translucent archwires from BioMers present a more esthetic option for patients, their mechanical response is less reliable than alloy wires, possibly compromising treatment efficiency.

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CHAPTER 1
INTRODUCTION

Fiber-reinforced composite has been used in dentistry for at least thirty years. It has been utilized in many areas: in prosthodontics for fixed partial dentures, in endodontics as posts and cores, in periodontics for periodontal splinting, and in oral surgery for trauma stabilization (Cacciafesta et al., 2008). With increasing esthetic demands, fiber-reinforced composite has also been used to replace alloy wires in orthodontics. Some passive applications are bonded lingual retainers and bonded pontics replacing missing lateral incisors (Burstone et al., 2000). As an active application, research has shown that fiber-reinforced composite can replace stainless steel wires to join segments of teeth together as an anchorage unit (Burstone et al., 2000; Cacciafesta et al., 2005). Taking active application one step further, fiber-reinforced composite archwires have been developed to be used in conjunction with ceramics and polycarbonates brackets to obtain ultimate esthetic results. They have been utilized in clinical trials (Chudasama & Jerrold, 2008). The translucent optical property of fiber-reinforced composite wires meets the esthetic demand of orthodontic patients. However, studies need to be conducted to see if it also has the desired mechanical properties for clinicians to utilize it in active orthodontic treatment.

Some research has been conducted on fiber-reinforced composite wires to compare their mechanical properties to alloy wires' mechanical properties (Chai et al., 2005; Fallis & Kusy, 2000; Imai et al., 1999; Jancar & Dibenedetto, 1993). Since composite archwires will be in the oral cavity for a substantial period of time, it is important to determine the effect water has on the wires. Many studies have shown that water storage decreases the strength of fiber-reinforced composite (Chai et al., 2005; Imai et al., 1999; Jancar & Dibenedetto, 1993). This hydrolytic degradation is due to water molecules diffusing into the resin matrix and acting as plasticizers which make the movement of resin polymer chains easier under stress (Chai et al., 2005). Many of these studies were done using prototypes; therefore, it is important to conduct a study on the effect of water storage on a commercially available fiber-reinforced composite wire.

The goal of this research was to study the effect of water storage on fiber-reinforced composite archwires from BioMers Products, LLC (Jacksonville, FL) and compare it to that of conventional nickel-titanium archwires (Nitinol Classic from 3M Unitek, Monrovia, CA), stainless steel wires, and beta-titanium wires (Beta III Titanium, 3M Unitek) using a three-point bend test.

CHAPTER 2
LITERATURE REVIEW

Orthodontic Wires

With the advancement of technology, many new materials have been introduced to the orthodontic armamentarium to help facilitate treatment. It is common in orthodontic practices to routinely use multiple types of archwires based on materials. The most common archwire materials currently are stainless steel, cobalt-chromium alloy, nickel-titanium alloys, and beta-titanium. In the past couple of decades, composite wires were introduced, investigated, and used clinically. Each of these materials mentioned above has indications for use in orthodontic treatment. The ideal properties of a wire for orthodontic usage should include high strength, high formability, a large range, and low stiffness for most applications (Proffit et al., 2007). An example of applications when a stiff wire is preferred will be utilizing sliding mechanics for space closure. The ideal wire should also be solderable and weldable for adding attachments, like hooks or spurs (Nikolai, 1997; Proffit et al., 2007). The material for an ideal wire should be able to undergo heat treatment to relieve the stress built up from bending or twisting (Nikolai, 1997). Along with these desired properties, the ideal wire should also have a reasonable cost (Proffit et al., 2007), be biocompatible, and demonstrate good esthetics.

Chronology of Alloy Orthodontic Wires

Precious alloy archwires

Before the development of archwire, in the late 1800, “arch bow” was used in orthodontic treatment. Arch bows were round wires with the dimension of 0.032 to 0.036 inches and was made from precious alloy like nickel-silver or platinum-gold alloy (Nikolai, 1997). The arch bow predated the invention of brackets and it was also referred to as an E-Arch, Edward Angle’s first appliance. The arch bow had threaded ends that passed through the tubes on bands, which were only on the terminal molars, and small nuts were placed either mesial or distal to the tube. These nuts could be activated to change the perimeter of the arch (Proffit et al., 2007). The arch bow could be expanded or constricted to control the transverse dimension of the arch form. Besides the terminal molars, all other teeth were ligated to the arch bow individually. Due to the stiffness of the arch bow from its diameter, precise movement of teeth or leveling of the arch was not possible with this appliance. Therefore, Angle developed ribbon arch, 0.020” x 0.050” gold wire (Nikolai, 1997). The ribbon arch was placed into the vertical slot behind the tube and held with pins. The springiness of this wire made it efficient in aligning teeth. However, the flexibility of the ribbon arch made it difficult to generate enough moments for torquing roots (Proffit et al., 2007).

In the 1920s, Edward Angle developed an edgewise appliance by reorienting the vertical slot into a horizontal slot, 0.022 by 0.028 inches. The rectangular wire was inserted into the slot 90 degrees to the orientation of the ribbon arch, so it was named an “edgewise” appliance (Nikolai, 1997; Proffit et al., 2007). The wings and the slot of the bracket provided control of tooth movement in all three planes of space. Initially, the wires made to be used with the edgewise appliance were still made of precious metal alloys. These gold- and silver-alloys were too soft in such a small dimension to achieve some stabilizing effects desired in orthodontic treatment (Nikolai, 1997).

Stainless steel wires

In the late 1920s, clinicians started incorporating the use of stainless steel in orthodontic treatment (Nikolai, 1997). Stainless steel contained a high content of chromium, which contributed to its corrosion resistance (Proffit et al., 2007). Comparing to precious metal alloys, this metal alloy had higher strength, springiness, ductility, stiffness, and corrosion resistance in the oral environment, and much lower cost. Therefore, stainless steel started replacing precious alloys in orthodontics even before the cost of the precious alloys started to become excessively expensive (Nikolai, 1997). Stainless steel is composed of mainly iron along with 17-20% chromium, 8-12% nickel, and up to 0.15% carbon. American Iron and Steel Institute types 302 and 304 austenitic stainless steel, also referred to as 18-8 stainless steel due to their percentages of chromium and nickel contents, are widely utilized in orthodontics. With all of its good mechanical properties, stainless steel has the disadvantages of high force delivery, low springiness, and susceptibility to intergranular corrosion after heating (O'Brien, 2008).

In the mid-1930s, multistrand stainless steel wires were evaluated for its usage in orthodontics via the labiolingual, twin wire technique (Nikolai, 1997). Multistrand wire is made from twisting small diameter wires together. For example, a multistrand wire made from twisting two 0.010 inch round wires has the springiness equivalent to a strand of 0.010 inch round wire, but its strength is twice as much as a strand of 0.010 inch round wire (Proffit et al., 2007). However, it is hard to place precise permanent bends, so it has never become the main working force in orthodontic archwires (Nikolai, 1997).

Cobalt-chromium wires

After World War II, cobalt-chromium alloy was first developed by Elgin Watch Company to replace watch mainsprings that were susceptible to corrosion. In the mid-1950s, cobalt-chromium wires started to be utilized for orthodontic purposes (Nikolai, 1997). These wires have many similar properties as stainless steel wires after it has been hardened by heat treatment, but they are supplied by the manufacturers in softened stages for clinicians to take advantage of its formability. Cobalt-chromium alloys are available commercially as Elgiloy (Rocky Mountain Orthodontics, Denver, CO), Azura (Ormco, Glendora, CA), and Multiphase (American Orthodontics, Sheboygan, WI) (Kapila & Sachdeva, 1989). Elgiloy, the most widely used, has a composition of 40% cobalt, 20% chromium, 15% nickel,15.8% iron, 7% molybdenum, 2% manganese, 0.16% carbon, and 0.04% beryllium (O'Brien, 2008). Elgiloy archwires are supplied in four tempers (resiliencies): soft (blue), ductile (yellow), semi-resilient (green), and resilient (red) (Kapila & Sachdeva, 1989). They are color coded for ease of differentiation for clinicians. Blue Elgiloy, being the least resilient among the four forms, is the most commonly used form and is preferred when extensive bending, soldering, or welding is required. It is also the wire of choice while making archwires with the multiloop edgewise archwires (MEAW) technique (Sasaguri, 2009). After heat treatment, the yield strength of blue Elgiloy can increase by 20% to 30%. Cobalt-chromium alloys have a slightly higher cost than stainless steel. They have very high formability comparing to stainless steel before heat treatment, but the elastic force delivery is comparable with stainless steel after being heat-treated (O'Brien, 2008). Cobalt-chromium wires have lower springback compared to stainless steel wires of the same size, but with proper heat treatment, the springback can be improved (Kapila & Sachdeva, 1989).

Nickel-titanium wires

In the early 1960s, a nickel-titanium (NiTi) alloy was developed initially for the space program at Naval Ordnance Laboratory (NOL); hence it was named nitinol. NiTi wires were introduced into orthodontic treatment in 1971 (Andreasen & Hilleman, 1971). The NiTi wires are composed of approximately 55% nickel and 45% titanium by weight (O'Brien, 2008).

Many metal alloys, including stainless steel and NiTi, can have more than one form of crystal structure. The martensitic form exists at lower temperatures while the austenitic form dominates at higher temperatures. For most of the metal alloys, including stainless steel, this phase transitional temperature is at hundreds of degrees. However, NiTi alloys utilize their low phase transition temperatures to obtain their two unique properties that are useful in orthodontics: shape memory and superelasticity (Proffit et al., 2007).

Shape memory refers to the ability of NiTi alloy to recover its original shape that is set at a temperature above the martensite-austenite transition temperature, after being plastically deformed in the martensitic phase. This can be accomplished through heating the deformed martensitic wire past the transition temperature; so this property is sometimes called thermoelasticity. Superelasticity is based on the change between austenite and martensite in response to applied force. This stressed-induced change from austenite to martensite is made possible due to its transition temperature being very close to room temperature (Proffit et al., 2007). Superelastic NiTi wires have a “plateau” region in its deactivation part of the stress-strain cycle (Nikolai, 1997). This shows the relatively constant force over a range of tooth movement during unloading. This is a highly desirable property for an orthodontic archwire.

In the late 1970s, Nitinol (Unitek, Monrovia, CA) became very popular due to its great property in springiness (Proffit et al., 2007). The original Nitinol wires did not have the shape-memory effect and were in stabilized martensitic form in the mouth (O’Brien, 2008; Proffit et al., 2007). However, it had two very important properties for orthodontic usage. The first one was a very low elastic modulus which decreased its force delivery to about one fifth of the force delivery for the stainless steel wires with the same cross-section dimension. The second useful property was the extremely wide elastic working range which allowed the wire to be inserted into brackets on malpositioned teeth without being permanently deformed nearly as much as stainless steel (O’Brien, 2008). Orthonol (Rocky Mountain), also a martensitic NiTi alloy, has similar properties to Nitinol, but it has better formability. This group of stabilized martensitic alloy is referred to as M-NiTi (Proffit et al., 2007).

In 1985, Burstone et al. reported a new form of NiTi that was developed in China (marketed as Ni-Ti by Ormco/Sybron) that was mostly in the form of body centered cubic austenite (Burstone et al., 1985; O’Brien, 2008). In 1986, Miura et al. described a Japanese NiTi alloy (Sentalloy by GAC International) wire that had similar properties as

Chinese NiTi (Miura et al., 1986). This group of wire is referred to as A-NiTi. These wires demonstrate an extraordinary property of NiTi alloys: superelasticity (Proffit et al., 2007).

The properties of A-NiTi, long range of activation with relatively constant force during deactivation, make it particularly useful as an initial archwire or as a coil spring. M-NiTi is valuable when the clinician desires a flexible but larger and stiffer wire during the later part of treatment. Therefore, smaller round NiTi is usually A-NiTi, while larger rectangular NiTi should be made from M-NiTi for better performance (Proffit et al., 2007).

Copper-nickel-titanium wires have the shape-memory behavior activated at body temperature. Copper Ni-Ti (Ormco) wires contain 5% copper. The three different types of Copper Ni-Ti wires achieve shape memory at 27°C, 35°C, and 40°C. The 27°C variant will be useful in mouth breathers, while 40°C variant is only activated when drinking hot fluids or eating hot food (O'Brien, 2008).

NiTi wires have the lowest force delivery among all orthodontic alloys. Other good properties of NiTi wires are superior springback, superelasticity, shape memory, and adjustable force delivery through heat-treatment of superelastic NiTi alloys. These qualities made NiTi wires very popular in orthodontics. However, there are some disadvantages associated with NiTi wires as well: they are more expensive, exhibit high friction, have lower corrosion resistance compared to other wires, are not solderable, and cannot withstand cold bending as well (Kapila & Sachdeva, 1989; Proffit et al., 2007; O'Brien, 2008; Valiathan & Dhar, 2006).

Beta-titanium wire

In the early 1980s, beta-titanium was developed primarily for orthodontics, after the introduction of Nitinol but before the advancement of A-NiTi. This alloy was first marketed by Ormco as TMA (titanium-molybdenum alloy) (Proffit et al., 2007) and contains 80% titanium, 11.5% molybdenum, 6% zirconium, and 4.5% tin (Kusy, 1997). Molybdenum is added to improve the formability of the wire (O'Brien, 2008). Overall, the properties of TMA wires are in between those of stainless steel and M-NiTi wires (Proffit et al., 2007). TMA wires have good combination of strength and springiness, and excellent formability. Therefore, it is a good choice for a finishing wire. Its disadvantages are high cost and high friction (O'Brien, 2008; Proffit et al., 2007).

Demand for Esthetic Appliances

The history of fixed appliances for orthodontics started with bands that had welded tubes and brackets. Banding every tooth for orthodontic treatment was necessary before bonding materials and techniques were improved. In the 1980s, direct bonding of brackets became routine for orthodontic treatment in fixed appliances (Proffit et al., 2007). This leap not only decreased the chair time of the clinician by not needing to sequentially band each tooth but also reduced the discomfort of the patient by not having separators between every tooth. This also significantly decreased the metallic appearance of the appliance; hence, it made fixed appliances more esthetic.

With the increased population of adults seeking orthodontic care in the past decades, the demand for more esthetic appliances also increased (Imai et al., 1999; Valiathan & Dhar, 2006). There was a trend for smaller metallic brackets in an effort to minimize a metallic appearance; however, its effect was very limited. Some clinicians considered lingual orthodontics to meet the esthetic needs of patients, but they discovered the additional technical difficulty and time requirement associated with working mainly on the lingual aspect of the teeth. The lingual appliances also have decreased efficiency in many areas compared to labial appliances due to the shorter interbracket distance. Clear aligners are very esthetic and comfortable for patients to use. However, many complex tooth movements are hard to achieve with using aligners (Russell, 2005). Therefore, clear brackets made from ceramics and polycarbonates became popular choices for clinicians and patients seeking a better esthetic solution for labial orthodontic appliances (Valiathan & Dhar, 2006).

Although clear brackets are very esthetic, they are still used with unesthetic metal archwires since alloys are currently the most effective materials in producing desired tooth movements (Burstone et al., 2011). Therefore, to achieve better esthetic results, research is done to produce an acceptable archwire material that combines the desired mechanical properties with esthetics. Some companies manufacture stainless steel and NiTi wires with tooth-colored polymers or inorganic coatings (Elayyan et al., 2010). For example, Marsenol is a NiTi wire coated with elastomeric poly tetra fluroethyl emulsion (ETE) and Lee White Wire is coated with a tooth colored epoxy coating (Agwarwal et al., 2011). Research needs to be done on the coated archwires to know how the coating affects the properties and dimension of the underlying wire. The durability of the coating needs to be improved because it can be easily damaged from forces of mastication and enzyme activities in the mouth (Kusy, 2002). Although the coating made the metallic wires much more esthetic, it is still not translucent, nor transparent. Therefore, some manufacturers developed composite wires as a new esthetic solution (Figure 1).

Figure 1. Different esthetic results with different combinations of wires and brackets. Upper arch: ceramic brackets with fiber-reinforced composite wire. Lower arch: ceramic brackets with alloy wire

Composite Wires

There are two classes of composite wires: self-reinforced and fiber-reinforced. The self-reinforced composite wire made from polyphenylene has high springback and ductility (Goldberg et al., 2011). It has high rigidity, strength, and hardness due to molecular-level reinforcement. This reinforcement is done by inserting flexible segments into a rigid all-phenylene chain to break it up into blocks of rigid phenylene groups, which reinforces itself, hence the name of self-reinforced polymer (Burstone et al., 2011). Burstone et al. (2011) reported that polyphenylenes “have increased hardness and resistance to stress relaxation. Along with good formability, torque control, and translucency, this thermoplastic polymer might be an efficient and esthetic labial orthodontic wire." The result from a clinical trial performed by New Ortho Polymers of Farmington, CT referenced by Goldberg et al. (2011) found that polyphenylene wires are efficient for moving teeth during the initial phase of orthodontic treatment.

Another way to enhance the properties of a polymeric wire is by fiber reinforcement (Burstone et al., 2011). With the addition of fibers, for example glass, the reinforced polymer increases in strength and rigidity which make it suitable for being used in many fields. This material was first utilized in sporting goods, like solid fiber-glass fishing rods. As the manufacturing techniques improved, the usage of fiber-reinforced composite has also extended to automotive, aerospace, electronic, and medical/dental fields (Fallis & Kusy, 2000). In 2001, over 3.3 billion pounds of polymeric composites were used in the United States (Lackey et al., 2003).

Starting from the 1960s, there were reports of denture resin reinforced with different types of fibers: glass, carbon, and aluminum and sapphire whiskers (Valiathan & Dhar, 2006). In the 1980s, many authors had recommended using fiber-reinforced polymer in denture base, prosthodontic frameworks for implants, splints (Goldberg & Burstone, 1992; Goldberg et al., 1994), and orthodontic retainers (Bearn, 1995). Because of its esthetics value, research has been done to evaluate the mechanical properties of unidirectional fiber-reinforced composite for active orthodontic treatment (Goldberg & Burstone, 1992; Jancar & Dibenedetto 1993; Jancar et al. 1993; Kennedy et al., 1998).

Manufacturing of continuous, fiber-reinforced composite

Some fiber-reinforced composite wires have been manufactured using a process called pultrusion. Pultrusion was developed by W. B. Goldsworthy in 1950 and is one of the oldest methods in manufacturing continuous, unidirectional fiber-reinforced composite (Kennedy & Kusy, 1995). As opposed to an extrusion process for manufacturing metal, pultrusion involves pulling the materials through the machine (Strong, 2008). A schematic diagram of the typical thermoset pultrusion process is shown in Figure 2. The process begins with pulling continuous, raw fibers from the reinforcement handling systems, also known as creels (Martin & Sumerak, 1987). The fibers pass through a thermoset resin bath and are pulled into a general shape on the preform station. The fully resin-impregnated fibers are pulled through a die that has the form of the final shape of the product. The die is heated which cures the resin. The hot, constant cross-section profile is pulled out of the mold and cools by ambient or forced air or water as it enters the pulling system, which provides the force to move materials through the entire system. Upon exit, the profile is cut to a desired length by an automatic cutting station (Martin & Sumerak, 1987; Strong, 2008). Pultrusion is a highly efficient, continuous process and has a high material utilization rate of more than 95% (Strong, 2008; Advani & Sozer, 2003). This means only 5% of the materials used during pultrusion manufacturing process is lost or wasted.

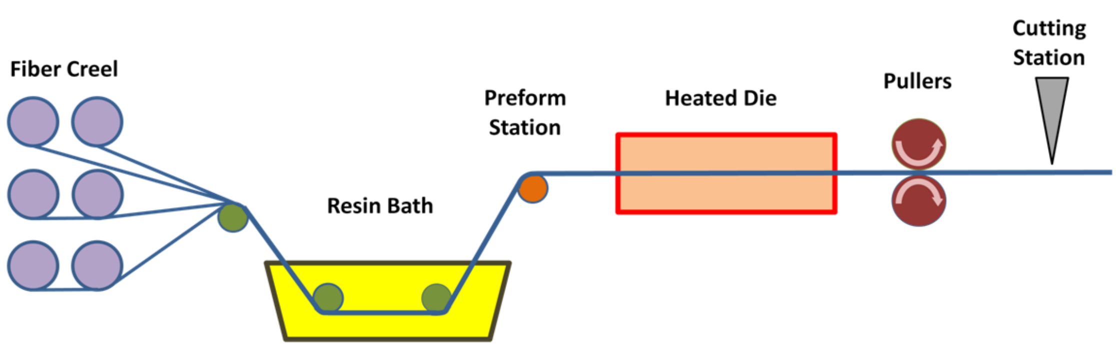

Figure 2. Schematic diagram of the typical thermoset pultrusion process.

The fiber-reinforced composite wires from BioMers are manufactured through pultrusion which uses a patented shrinkable and flexible die that reacts to heat. In lieu of using the thermoset resin described previously, a photo-cured resin is used. The process is oriented vertically instead of horizontally. After the resin-impregnated fibers enter the die, the die is heated from top to bottom for it to shrink evenly to produce a uniform cross-section profile. Then a desired length of die with the profile is cut and bent along its length into an arch form. This section of profile is cured by UV light. Then the die is peeled from the fiber-reinforced composite (Gopal et al., 2005). This photo-curing process can manufacture highly preformed fiber-reinforced composite with a small profile (Lacky et al., 2003).

The effect of water storage on fiber-reinforced composite

Since orthodontic wires are utilized in the oral cavity, it is important to know the mechanical properties of the wires after they have been immersed in water for a period of time. Research has shown that water storage decreases the strength of composite (Chai et al., 2005; Imai et al., 1999). When fiber-reinforced composite is stored in water, water molecules diffuse into the resin matrix and act as a dispersant to increase the plasticity or fluidity of resin polymer chains; therefore, the strength of the composite decreases. Water sorption usually increases as the percentage of polymer matrix increases. Thus, the lower the percent fiber content of the fiber-reinforced composite, the more that composite absorbs water and is adversely affected by it (Chai et al., 2005).

CHAPTER 3
MATERIALS AND METHODS

Align A, B, and C, and TorQ A and B from BioMers Products (Table 1), 0.014”, 0.016”, 0.018”, and 0.019” x 0.025” NiTi (Nitinol Classic, 3M Unitek), 0.016” stainless steel (3M Unitek), and 0.019” x 0.025” beta-titanium (Beta III Titanium, 3M Unitek) archwires were tested in this study. Each type and size of archwires consisted of 10 specimens (n=10/type/size/condition).

Table 1. Specifications of BioMers fiber-reinforced composite wires from the manufacturer.

Wire Type	Dimension (inches)	Dimension (mm)
Align A	0.018	0.457
Align B	0.019	0.483
Align C	0.021	0.533
TorQ A	0.019 x 0.025	0.483 x 0.635
TorQ B	0.021 x 0.025	0.533 x 0.635

A 20 mm segment was cut with pliers from each end of the archwire. The diameter of the wire segments was measured at three different points on the wire using a digital caliber. A segment from one end of the archwire was stored in distilled water at 37°C for 30 days (Figure 3) while the other segment from the same archwire was stored dry. The segments were tested using 3-point bending at 37°C ± 2°C. The specimens were centered between the two support beams, which had a span length of 14 mm. The load was applied vertically with a universal testing machine (Model 5500R, Instron, Norwood, MA) (Figures 4 and 5) to the middle of the specimens at the rate of 2 mm per min to a maximum deflection of 3.1 mm, and then it was returned to its starting position at the same rate (Figure 6). The 3-point bend test was carried out following American Dental Association Specification No. 32 for Orthodontic Wires (ADA, 2000) with the modification that the support length was 14 mm instead of 12 mm. The modification is due to limitation of the fixtures. Also, Nakano et al. (1999) recommended using 14 mm because it is the average distance between the labial center of a mandibular lateral incisor and a first premolar on the same side of the arch.

The force required to deflect the specimens was monitored and recorded by dedicated software (Merlin, Instron) during loading (activation)/unloading (deactivation). Due to the curvature in the posterior segment of the fiber-reinforced composite wires, all the rectangular wires were tested edge-wise to prevent the wires from slipping off the testing fixture.

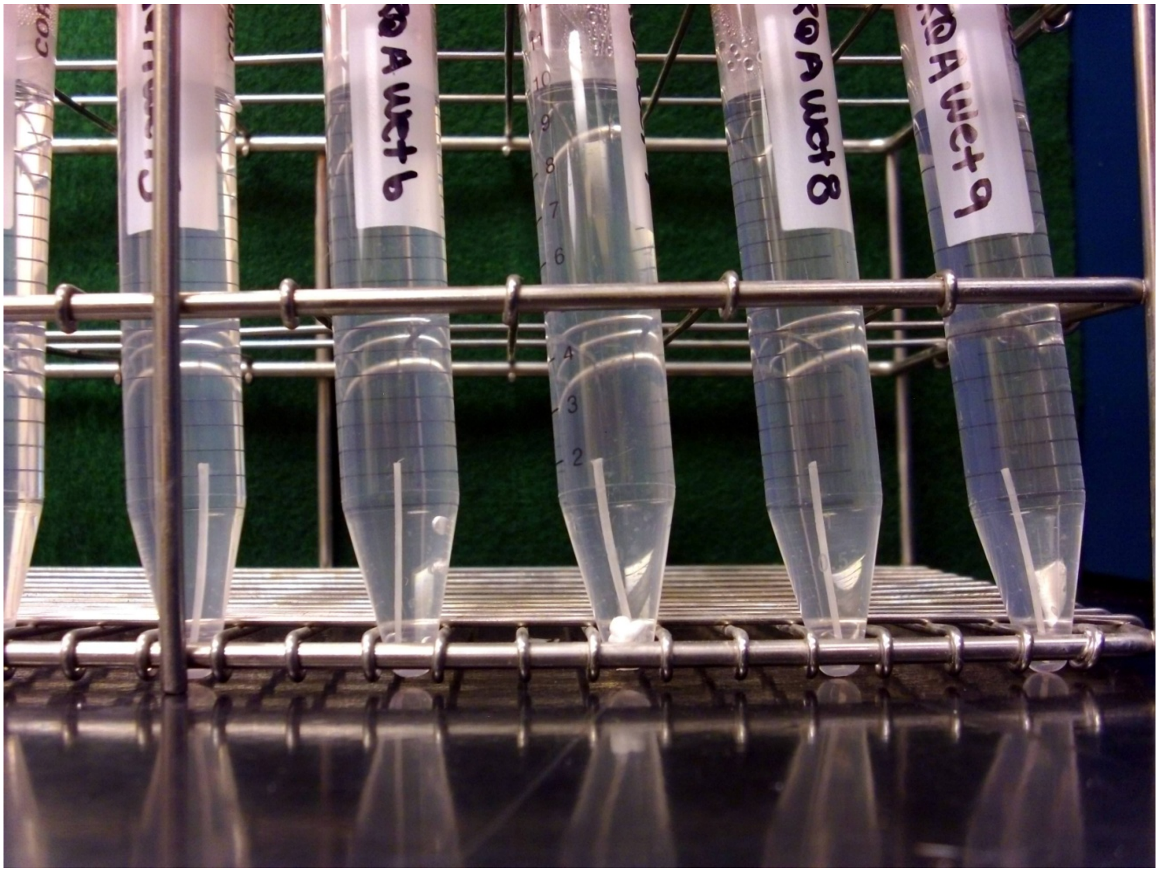

Figure 3. Wire segments stored in distilled water.

Figure 4. Instron 5500R machine for 3-point bending test.

Figure 5. Instron 5500R with the testing platform.
The temperature is controlled by the portable heater in the back and monitored with the probe attached to the platform.

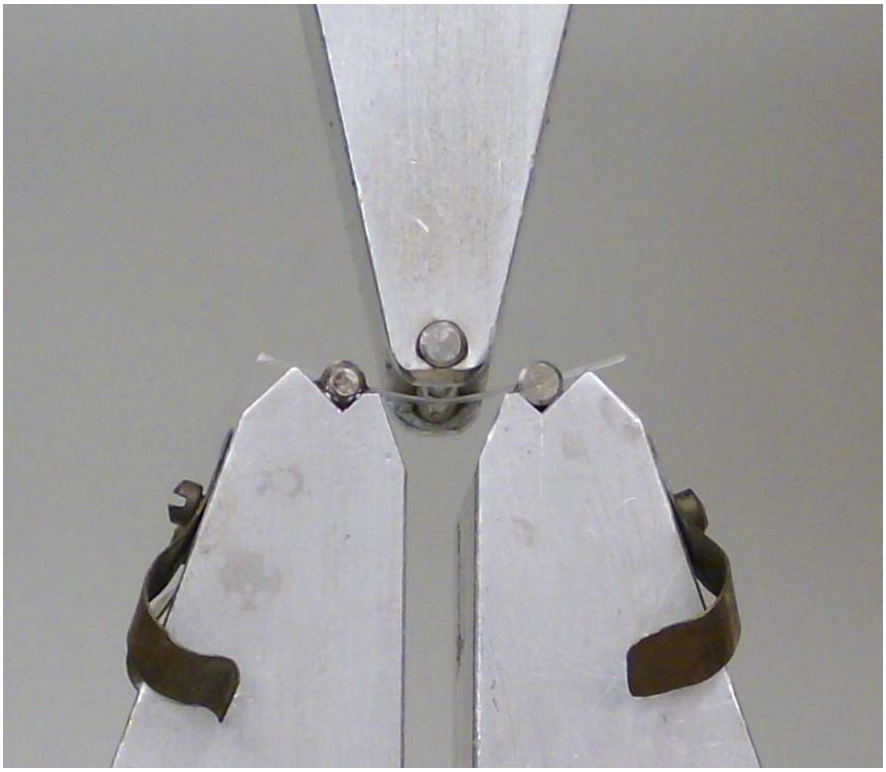

Figure 6. Three-point bending of a fiber-reinforced composite wire.

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 deflection during both activation and deactivation comprised the data harvested from each test. Additionally, the slope was converted to bending modulus (GPa) and the % elastic recovery computed. Statistical analysis was performed using a 2-way analysis of variance (ANOVA) with wire and condition (dry/wet) as factors followed by a post-hoc Tukey test when indicated. To compare parameters between the dry/wet conditions, a paired t-test was performed. All statistical tests were done using a $P < 0.05$ level of significance and statistical software (SPSS Inc., Chicago, IL).

CHAPTER 4
RESULTS

All the wire segments were measured at three different points along the segments; the averages of the measurements are listed in Table 2. The average dimensions of all wires were different from the dimensions specified by manufacturers. For each type or size of the alloy archwires (Nitinol Classic 0.014”, 0.016”, 0.018”, 0.019” x 0.025”; stainless steel 0.016”; beta-titanium 0.019” x 0.025”), no variation in dimensions were detected along the same wire segment nor among different specimens. In contrast to the alloy wires, the measurements taken from the BioMers fiber-reinforced composite wires (Align A, B, and C; TorQ A, and B) varied among the same type of specimens and also from one point of the same segment to another. The standard deviations are provided to demonstrate the variability. The measured dimensions, instead of manufacturer-specified dimensions, were used for calculating bending modulus.

Table 2. Dimensions of the wire specified by manufacturers comparing to the average dimensions measured by researcher at three different points of the wire segments.

Wire Type	Dimension specified by manufacturer (inches)	Dimension specified by manufacturer (mm)	Average dimension measured by researcher (mm)
Align A	0.018	0.457	0.456 ± 0.010 (1 SD)
Align B	0.019	0.483	0.468 ± 0.016 (1 SD)
Align C	0.021	0.533	0.524 ± 0.010 (1 SD)
Nitinol Classic 0.014"	0.014	0.356	0.340
Nitinol Classic 0.016"	0.016	0.406	0.390
Nitinol Classic 0.018"	0.018	0.457	0.435
Stainless Steel 0.016"	0.016	0.406	0.390
TorQ A	0.019 x 0.025	0.483 x 0.635	0.520 ± 0.014 (1 SD) x 0.720 ± 0.029 (1 SD)
TorQ B	0.021 x 0.025	0.533 x 0.635	0.590 ± 0.032 (1 SD) x 0.770 ± 0.035 (1 SD)
Nitinol Classic 0.019"x0.025"	0.019 x 0.025	0.483 x 0.635	0.470 x 0.630
Beta-titanium 0.019"x0.025"	0.019 x 0.025	0.483 x 0.635	0.470 x 0.630

The force and deflection were monitored during activation and deactivation of the round wires (Figures 7-13). Both Align A groups, stored dry or stored in water, had roughly the same activation and deactivation curves, and they both had a very small amount of permanent deformation as shown by the force level reaching zero when the wires were still deflected during deactivation (Figure 7). For Align B (Figure 8), the stored dry group had force-deflection curves similar in shape to that of the Align A groups. However, for the Align B stored in water group, there was a higher force delivery for activation comparing to its stored dry group, but due to a large drop of the force level at greater deflections (3.1 mm in Figure 8), the deactivation forces were less. Its permanent deformation was greater than that of the stored dry group.

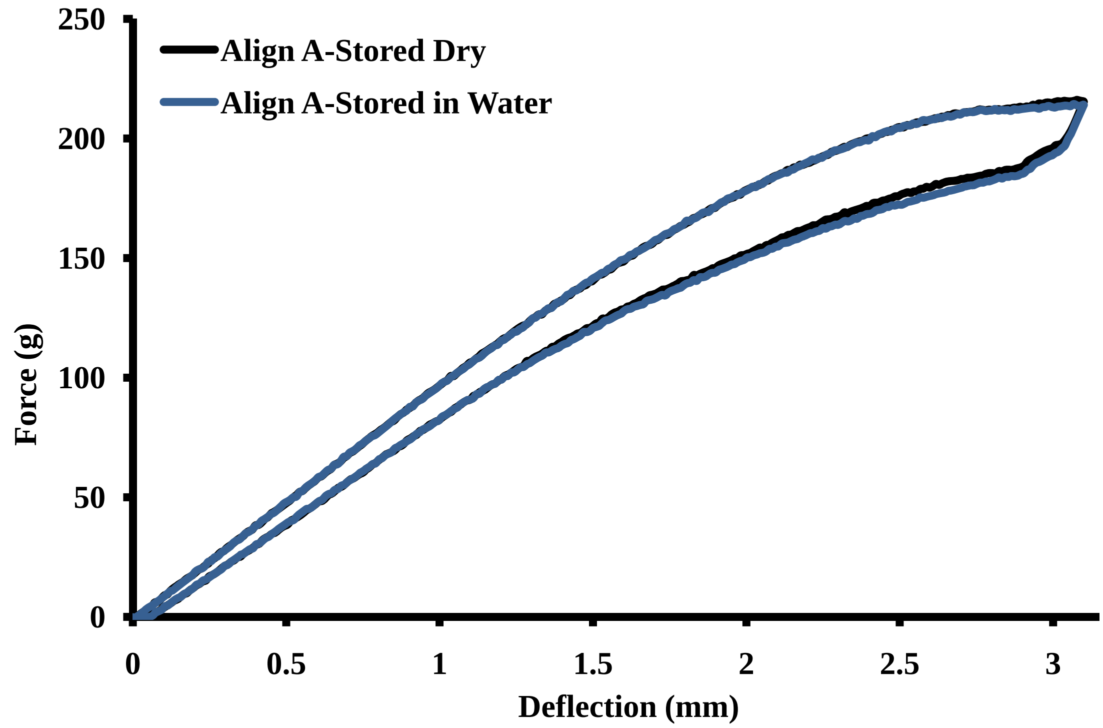

Figure 7. Graph example of force-deflection curve for Align A.

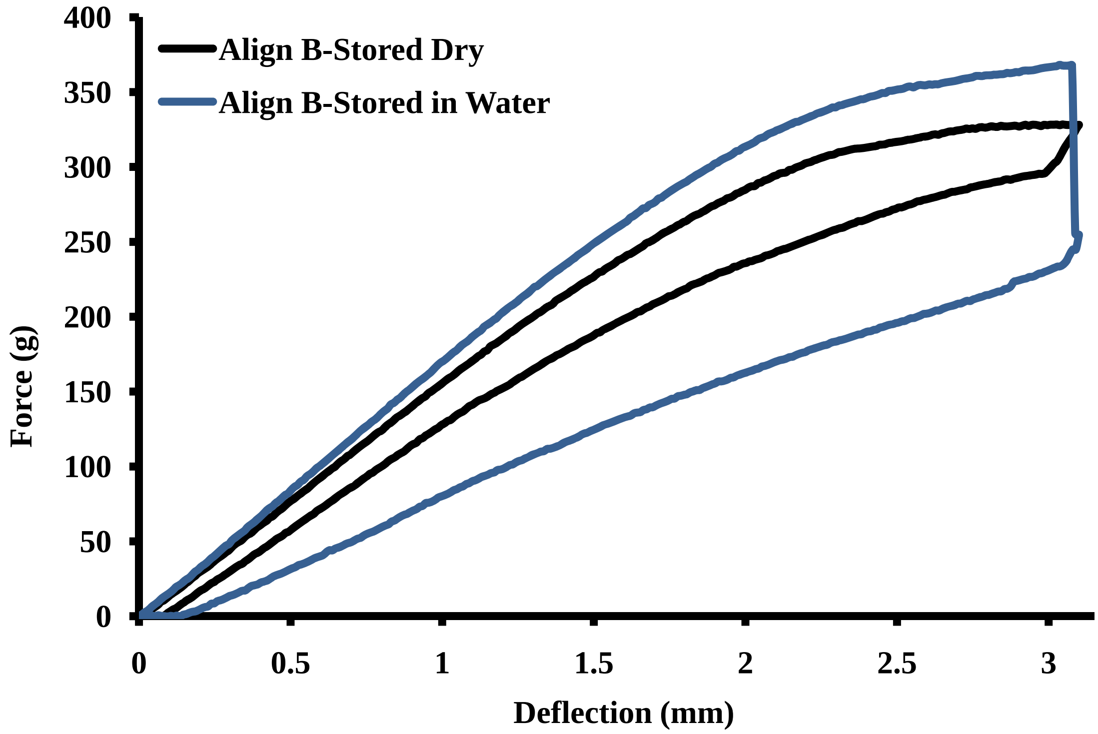

Figure 8. Graph example of force-deflection curve for Align B.

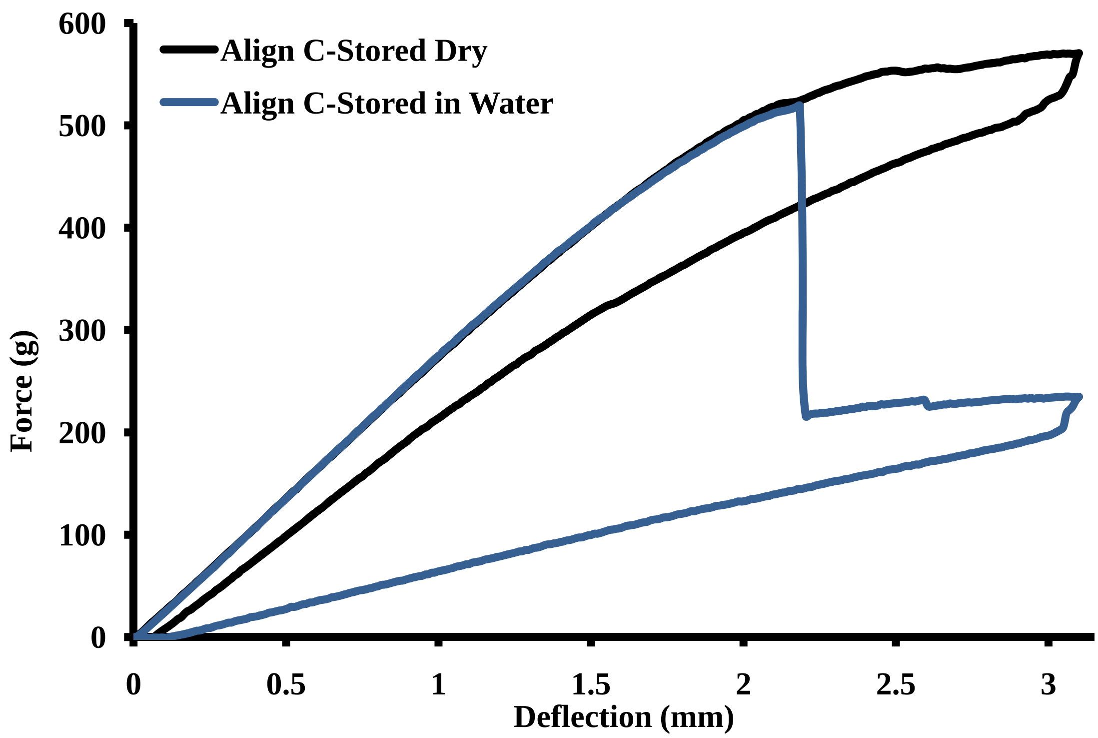

Figure 9. Graph example of force-deflection curve for Align C.

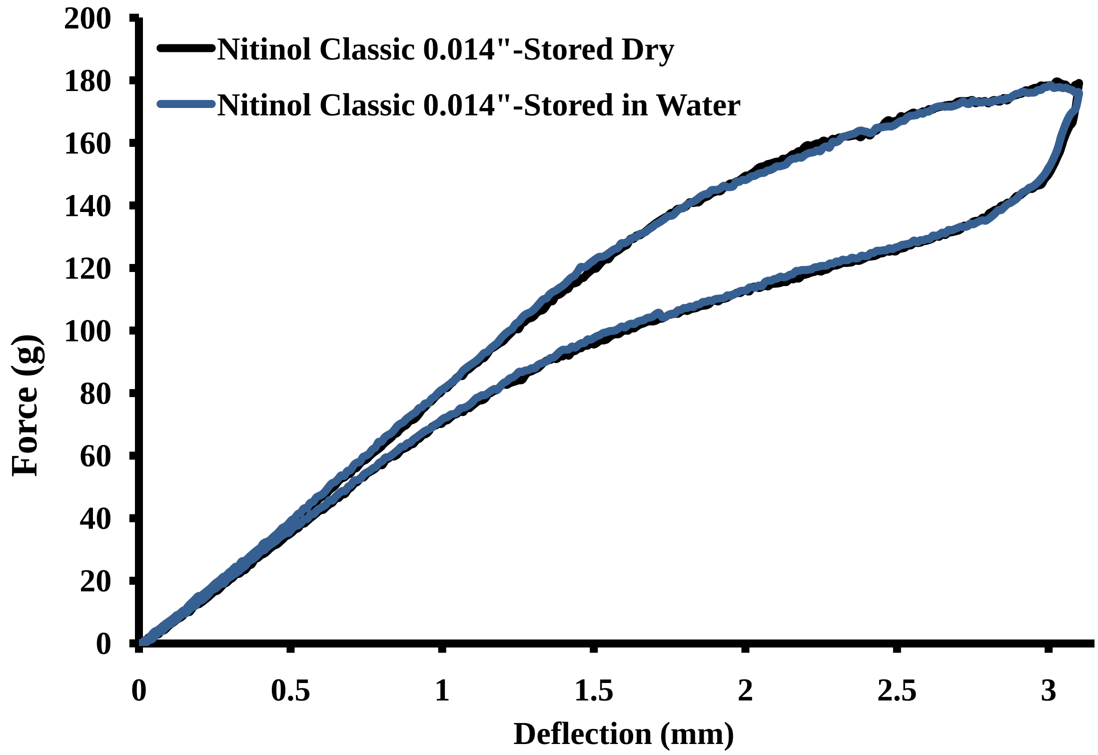

Figure 10. Graph example of force-deflection curve for Nitinol Classic 0.014”.

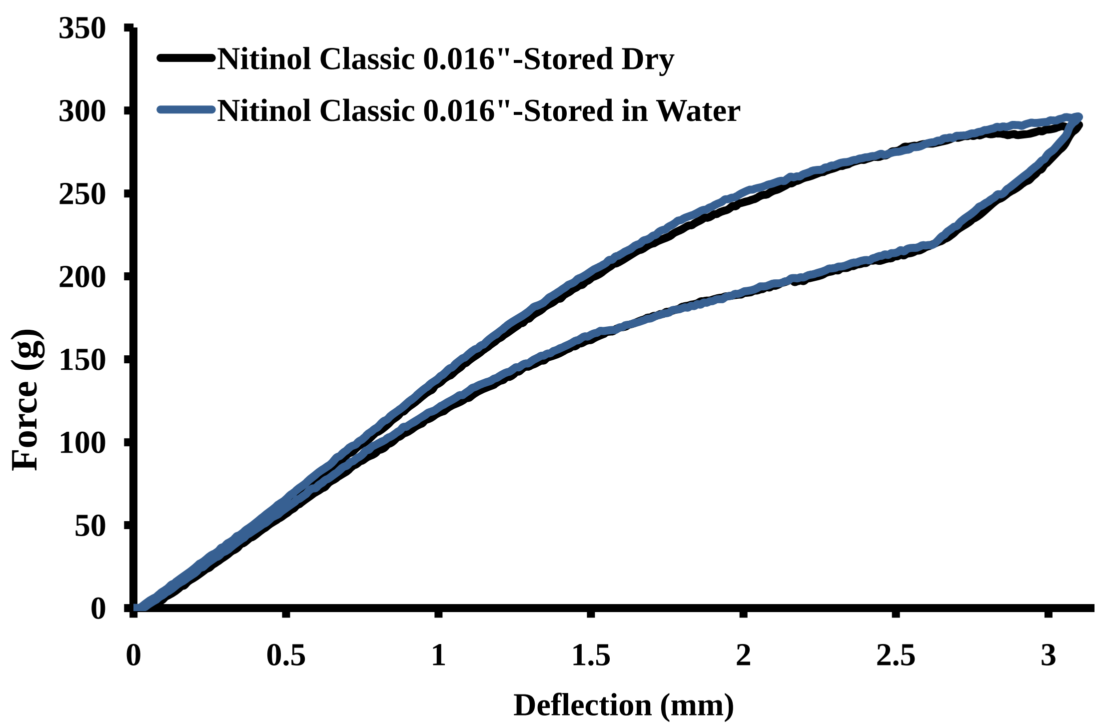

Figure 11. Graph example of force-deflection curve for Nitinol Classic 0.016”.

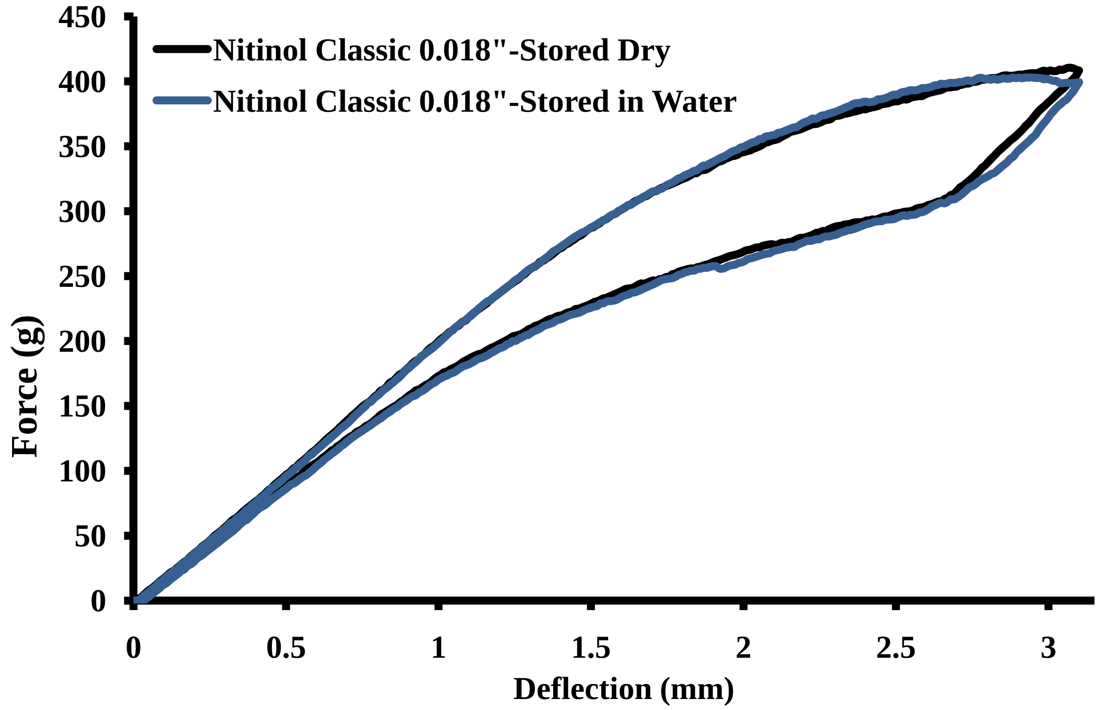

Figure 12. Graph example of force-deflection curve for Nitinol Classic 0.018”.

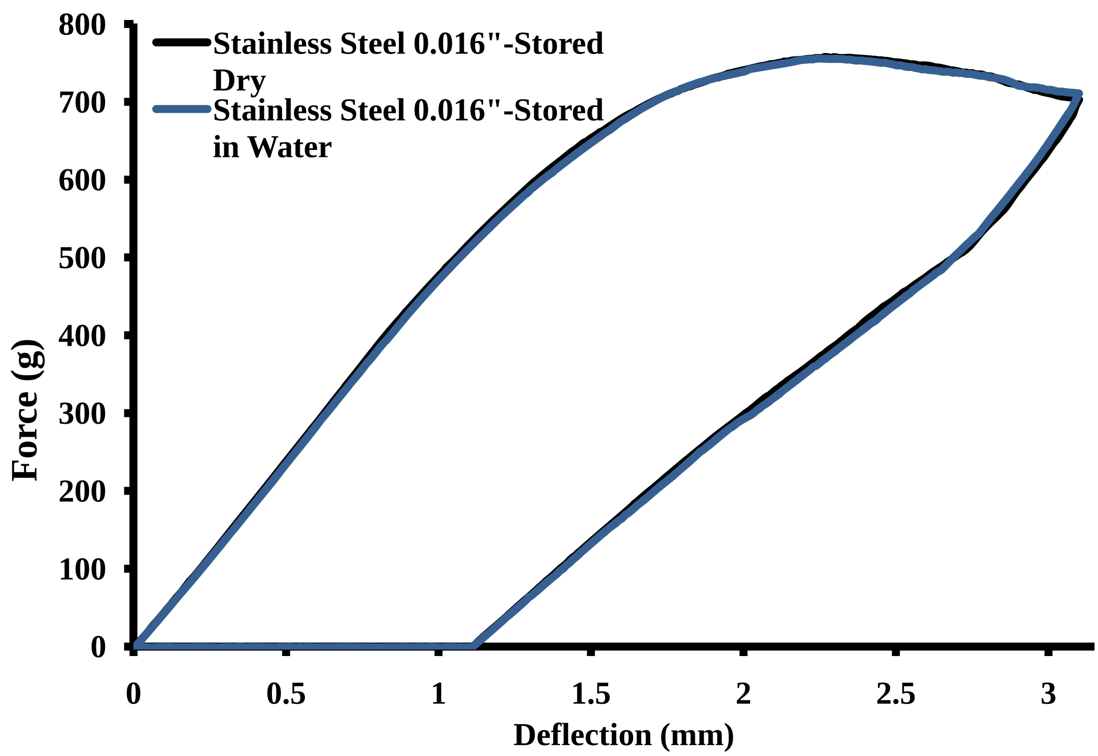

Figure 13. Graph example of force-deflection curve for stainless steel 0.016”.

The Align C stored dry group (Figure 9) had a similarly shaped force-deflection curve compared to the stored dry groups in Align A and B; however, its activation part of the curve was not as smooth as the other two groups when the deflection became larger. The Align C stored in water group had a large drop during activation like the Align B stored in water group, but this drop happened at a smaller deflection. The Align C stored in water group also had a bigger permanent deformation compared to its corresponding stored dry group.

The drops of force level observed in the stored in water groups for Align B and C (Figures 8 and 9) and the lack of smoothness in the activation part of the curve for Align C stored dry group were due to cracking or crazing (Figure 14). A crack is "an actual separation of material, visible on opposite surfaces of the part, and extending through the thickness. A fracture" (Pebly, 1987). Crazing is a "region of ultrafine cracks, which may extend in a network on or under the surface of a resin or plastic material. May appear as a white band" (Pebly, 1987). Due to the damage of the wires, the wires exerted less force compared to before crazing and cracking.

Nitinol Classic 0.014", 0.016", and 0.018" (Figures 10-12) had similar force-deflection curves. For all three groups, the stored dry groups and their corresponding stored in water groups generally had the same activation and deactivation curves. All Nitinol Classic wires exhibited little to no deformation. For stainless steel 0.016" (Figure 13), both stored dry and stored in water groups had about the same force-deflection curves. During activation, the force level actually started decreasing when the deflection was more than 2.3 mm. The permanent deformation of the stainless steel wires was also greater than all the other types of round wires.

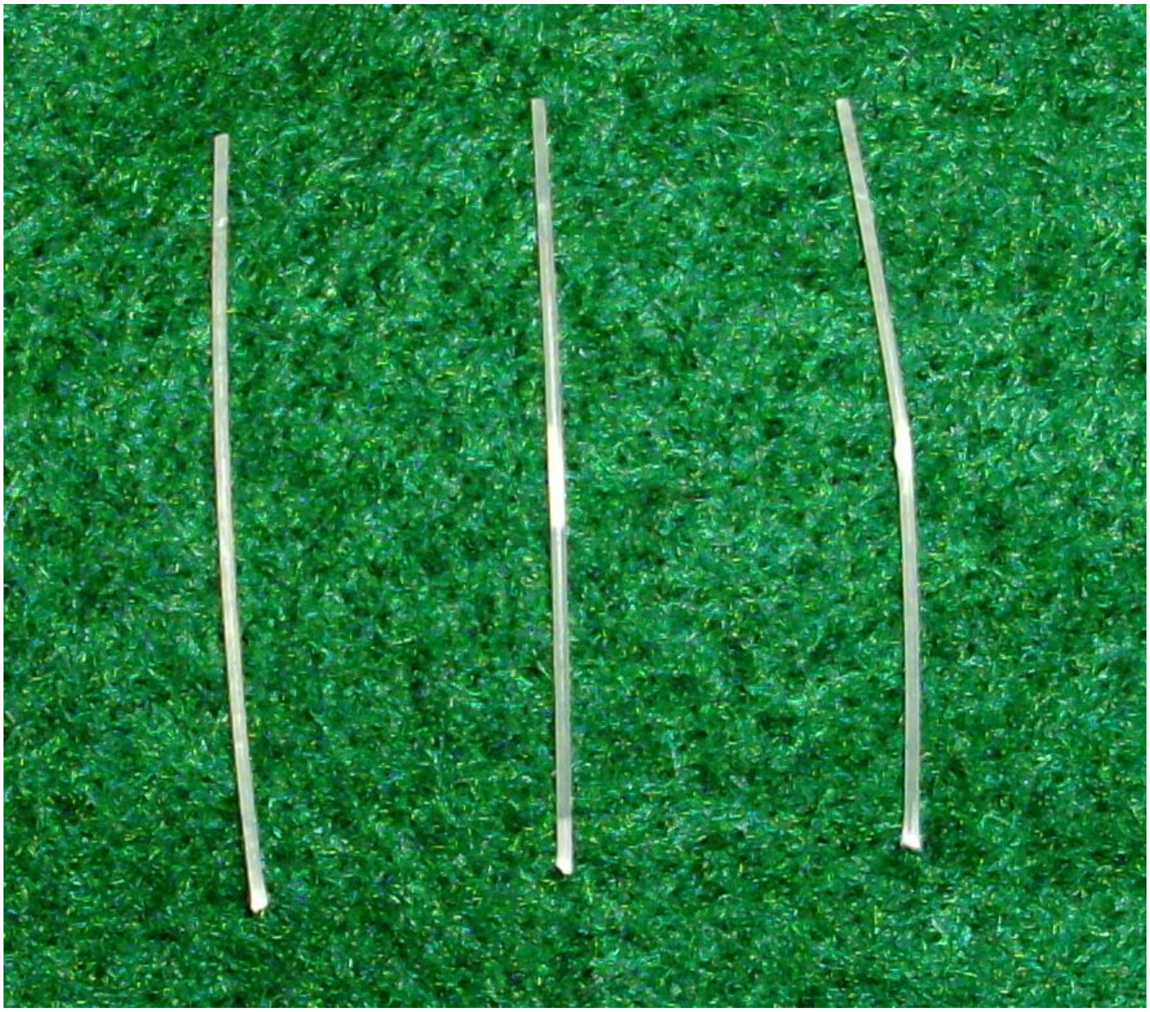

Figure 14. Crazing and cracking of the fiber-reinforced composite wires.
The wire on the left had wide spread minor crazing in its middle third section. The wire in the middle had a well-defined white band due to crazing, but the wire still remained structurally intact. The wire on the right had a well defined region of crazing, and it also was cracked in the middle of the white band. The white areas at the bottom end of all three wires were from cutting the segments from the archwires using pliers.

A comparison of round wires was done separately for stored dry groups and stored in water groups (Figures 15 and 16). For the stored dry group (Figure 15), all wires had the same general shape of the force-deflection curves, except stainless steel 0.016” wires which was excluded from the graph. In terms of stiffness and force delivery during activation, the wires were observed to follow the descending order: stainless steel

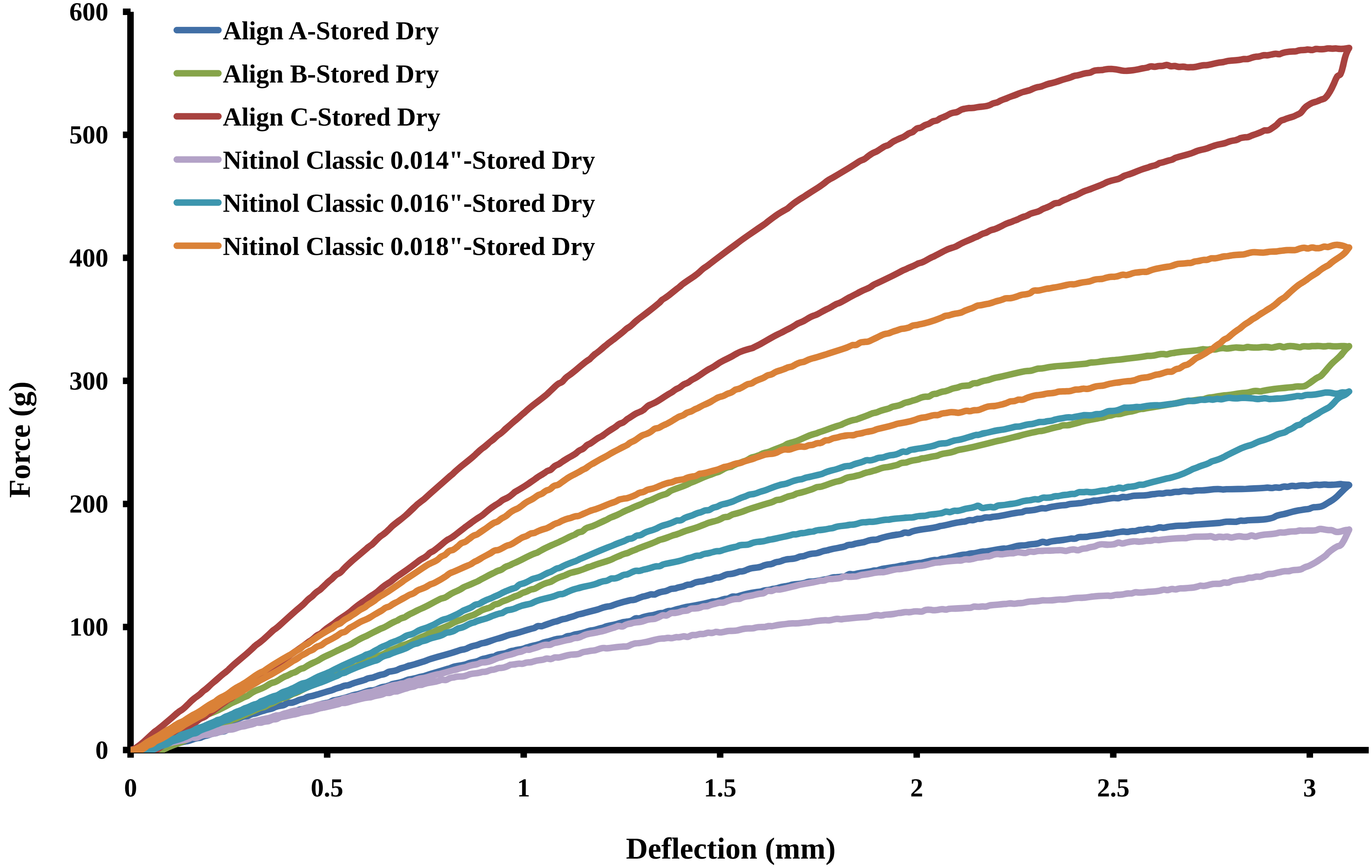

Figure 15. Comparison of force-deflection curves of round wires in the stored dry groups (stainless steel wires were excluded).

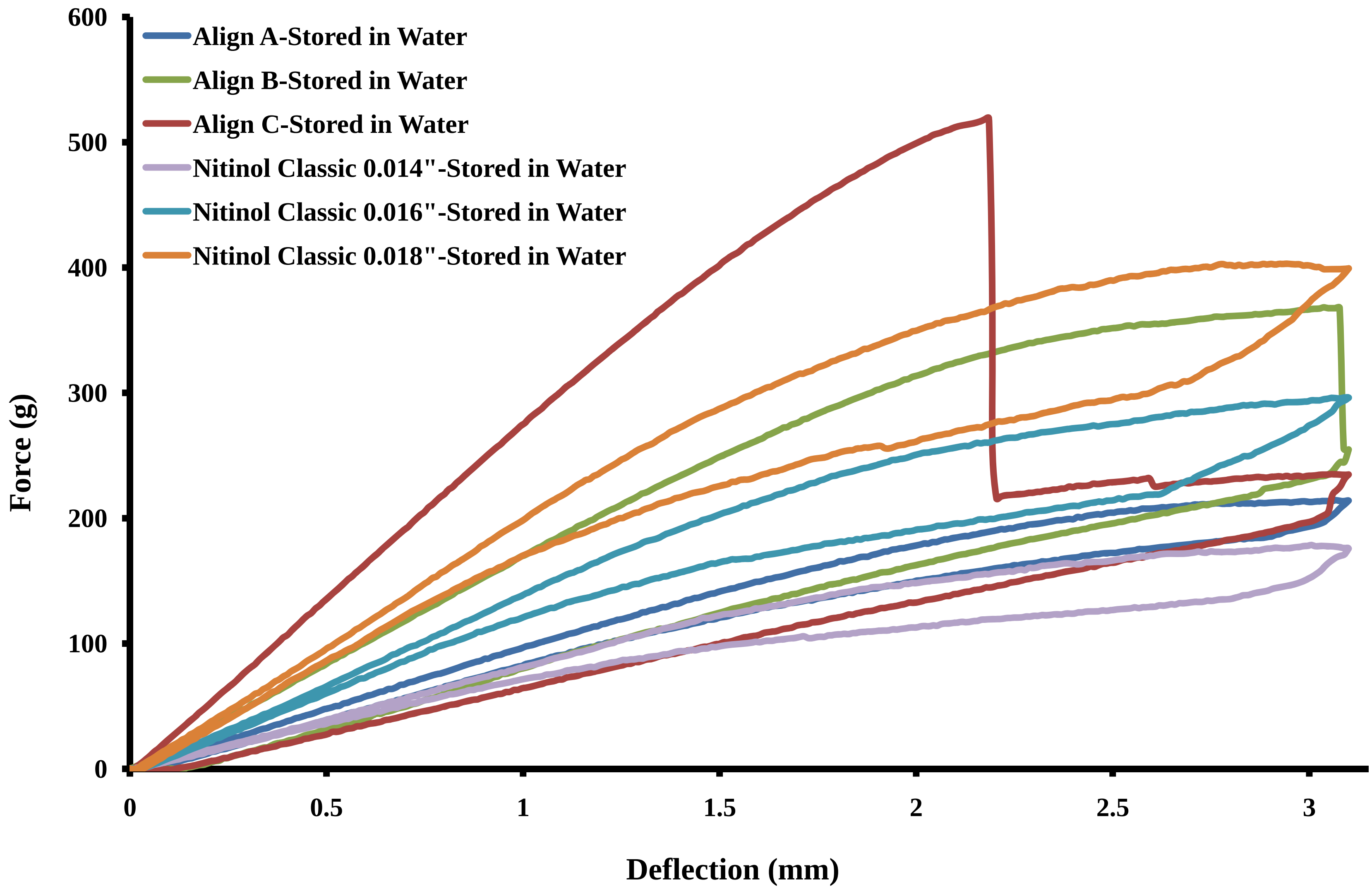

Figure 16. Comparison of force-deflection curves of round wires in the stored in water groups (stainless steel wires were excluded).

0.016”, Align C, NiTi 0.018”, Align B, NiTi 0.016”, Align A, and NiTi 0.014”. Multiple small drops in force levels were seen in Align C. For stored in water groups (Figure 16), the curves were similar to the ones in the stored dry groups, except Align B and C. Due to the cracking/crazing, there were sudden drops of force delivery, which made the subsequent activation and deactivation force levels much smaller. Another interesting finding was although the dimension of Align A was similar to Nitinol Classic 0.018”, its stiffness and force delivery were between that of Nitinol Classic 0.014” and 0.016”.

The bending values during activation and deactivation for round wires are shown in Tables 3 and 4. During activation for round wires, 0.016” stainless steel had the highest stiffness, bending modulus, and force delivery. The round wires listed in decending order of stiffness during activation are: 0.016” stainless steel, Align C, 0.018” Nitinol Classic, Align B, 0.016” Nitinol Classic, Align A, and 0.014” Nitinol Classic, which were all significantly different from each other. The bending moduli of the alloy wires were significantly greater than the fiber-reinforced composite wires. The force delivery levels at 1 mm and 2 mm deflections corresponded with the stiffness very well. However, for force levels at a deflection of 3 mm, the order had changed due to cracking of the fiber-reinforced composite wires. The cracking rates generally increased with the size of the fiber-reinforced composite wires; cracking rates for stored in water groups were higher than their corresponding stored dry groups. Due to cracking, the force delivery levels for Align C were significantly higher in stored dry group comparing to stored in water group. None of the alloy archwires cracked during the bending test.

Table 3. Bending values during activation for round wires.

ARCHWIRE	ACTIVATION					
	Stiffness (g/mm)	Modulus (GPa)	Force at 1 mm (g)	Force at 2 mm (g)	Force at 3 mm (g)	% with cracks (at deflection)
Align A (0.018”)-Dry	111 ± 18 F	28.9 ± 4.8 E	110 ± 17 F	199 ± 28 F	237 ± 29 DE	0
Align A (0.018”)-Water 30 d,37°C	117 ± 17 F	31.8 ± 5.8 E	115 ± 16 F	192 ± 18 F	231 ± 24 DE	30 (1.39 ± 0.34 mm)
Align B (0.019”)-Dry	172 ± 23 D	41.5 ± 3.5 D	169 ± 23 D	284 ± 93 D	298 ± 119 CD	50 (2.59 ± 0.56 mm)
Align B (0.019”)-Water 30 d,37°C	176 ± 13 D	41.1 ± 3.2 D	173 ± 13 D	317 ± 23 D	214 ± 154 CD	60 (2.60 ± 0.46 mm)
Align C (0.021”)-Dry	268 ± 13 B	39.7 ± 2.7 D	265 ± 14 B	478 ± 24 B	475 ± 151 C*	40 (2.55 ± 0.43 mm)
Align C (0.021”)-Water 30 d,37°C	258 ± 26 B	40.1 ± 2.9 D	254 ± 25 B	409 ± 133 B	163 ± 174 C*	100 (2.22 ± 0.44 mm)
Nitinol Classic 0.014”-Dry	82 ± 1 G	69.9 ± 1.1 B	81 ± 1 G	147 ± 2 G	175 ± 3 E	0
Nitinol Classic 0.014”-Water 30 d,37°C	82 ± 1 G	69.6 ± 0.8 B	81 ± 1 G	148 ± 2 G	175 ± 3 E	0
Nitinol Classic 0.016”-Dry	143 ± 1 E	70.5 ± 0.6 B	140 ± 1 E	249 ± 2 E	294 ± 4 CD	0
Nitinol Classic 0.016”-Water 30 d,37°C	143 ± 1 E	70.5 ± 0.7 B	140 ± 2 E	249 ± 2 E	292 ± 4 CD	0
Nitinol Classic 0.018”-Dry	205 ± 3 C	65.3 ± 0.8 C	201 ± 2 C	348 ± 4 C	406 ± 10 B	0
Nitinol Classic 0.018”-Water 30 d,37°C	205 ± 1 C	65.2 ± 0.4 C	201 ± 2 C	348 ± 3 C	408 ± 10 B	0
Stainless Steel 0.016”-Dry	489 ± 7 A	241.5 ± 3.5 A	475 ± 5 A	741 ± 4 A	717 ± 7 A	0
Stainless Steel 0.016”-Water 30 d,37°C	488 ± 3 A	240.5 ± 1.6 A	474 ± 5 A	739 ± 4 A	717 ± 9 A	0

Within each parameter, different letters denote significant differences ($p<0.05$) exist between types of wire.
* indicates a significant difference between dry and water-stored wires of the same type/size.

Table 4. Bending values during deactivation for round wires.

ARCHWIRE	DEACTIVATION					
	Stiffness (g/mm)	Modulus (GPa)	Force at 3 mm (g)	Force at 2 mm (g)	Force at 1 mm (g)	Elastic Recovery (%)
Align A (0.018")-Dry	98 ± 16 DE	25.7 ± 4.2 C	217 ± 27 CD	170 ± 24 B	94 ± 16 C	99.0 ± 0.7 AB
Align A (0.018")-Water 30 d,37^{o}C	90 ± 15 DE	24.0 ± 3.9 C	196 ± 54 CD	161 ± 19 B	86 ± 15 C	97.4 ± 4.6 AB
Align B (0.019")-Dry	110 ± 59 DE	26.8 ± 14.7 CD	257 ± 123 C*	191 ± 99 BC	103 ± 56 C	98.0 ± 2.8 B*
Align B (0.019")-Water 30 d,37^{o}C	72 ± 57 DE	16.0 ± 12.1 CD	175 ± 131 C*	123 ± 100 BC	64 ± 56 C	96.5 ± 3.3 B*
Align C (0.021")-Dry	176 ± 75 CD*	26.3 ± 11.2 D*	425 ± 147 C*	301 ± 125 B*	162 ± 72 BC*	96.7 ± 3.7 C*
Align C (0.021")-Water 30 d,37^{o}C	36 ± 36 CD*	5.6 ± 5.3 D*	124 ± 129 C*	65 ± 63 B*	30 ± 29 BC*	89.1 ± 4.9 C*
Nitinol Classic 0.014"-Dry	74 ± 1 E	62.9 ± 1.0 B	151 ± 2 D	114 ± 2 C	71 ± 1 C	99.9 ± 0.2 A*
Nitinol Classic 0.014"-Water 30 d,37^{o}C	74 ± 2 E	63.3 ± 2.1 B	150 ± 2 D	113 ± 1 C	71 ± 1 C	99.6 ± 0.3 A*
Nitinol Classic 0.016"-Dry	127 ± 2 C	62.6 ± 0.8 B	273 ± 3 C	193 ± 2 B	120 ± 2 B	99.4 ± 0.5 AB
Nitinol Classic 0.016"-Water 30 d,37^{o}C	128 ± 2 C	63.1 ± 0.8 B	271 ± 4 C	192 ± 2 B	121 ± 2 B	99.3 ± 0.6 AB
Nitinol Classic 0.018"-Dry	182 ± 2 B	58.2 ± 0.5 B	381 ± 13 B	265 ± 3 A	170 ± 3 A	99.2 ± 0.6 AB
Nitinol Classic 0.018"-Water 30 d,37^{o}C	182 ± 2 B	57.9 ± 0.6 B	382 ± 11 B	265 ± 3 A	170 ± 3 A	99.4 ± 0.3 AB
Stainless Steel 0.016"-Dry	320 ± 4 A	158.1 ± 2.1 A	652 ± 10 A	295 ± 6 A	0	64.4 ± 0.6 D
Stainless Steel 0.016"-Water 30 d,37^{o}C	320 ± 5 A	158.1 ± 2.3 A	650 ± 11 A	294 ± 6 A	0	64.5 ± 0.4 D

Within each parameter, different letters denote significant differences ($p<0.05$) exist between types of wire.
* indicates a significant difference between dry and water-stored wires of the same type/size.

For deactivation of round wires (Table 4), 0.016” stainless steel still had the highest stiffness. The stiffness of alloy wires followed the same trend as activation. However, due to the cracking of wires, fiber-reinforced composite wires all had lower stiffness values. Alloy wires still had higher bending moduli than the composite wires. The force delivery at 1 mm was 0 g for 0.016” stainless steel wires because the wires were significantly bent with the lowest elastic recovery. The average elastic recovery for all Nitinol Classic and Align A stored dry wires were above 99%. For Align C, all values between stored dry and stored in water groups were significantly different due to cracking in the stored in water group. Elastic recovery figures for fiber-reinforced composite wire stored in water groups were lower than their corresponding store dry group.

The graph examples for the rectangular wires are shown in Figures 17-20. For both groups of TorQ A, the wires started crazing or cracking when the activation deflection reached 1 to 1.5 mm (Figure 17). The wires started cracking again after 2 mm of deflection. Overall, the TorQ A stored dry group crazed or cracked less than the stored in water group. The same trend was detected in TorQ B wires (Figure 18). Both groups of TorQ B wires cracked or crazed multiple times during activation. The force delivery for the stored dry group was higher than that of the stored in water group. Nitinol Classic 0.019” x 0.025” groups behaved like round Nitinol Classic wires (Figure 19); force-deflection curves for both stored dry and stored in water groups followed each other closely with little deformation. However, Nitinol Classic 0.019” x 0.025” wires had the tendency to flip from edgewise to flat-wise. Therefore, their tests were terminated once the wires had flipped. Beta-titanium 0.019” x 0.025” groups performed like stainless steel wires, both stored dry and stored in water groups followed the similar force-deflection curves (Figure 20). At around 2.3 mm of deflection during activation, the force delivery decreased as the deflection increased. Large deformations were detected.

The comparisons of rectangular wires were done separately for stored dry groups and stored in water groups (Figures 21 and 22). For both storing methods, beta-titanium 0.019" x 0.025" wires had the highest force delivery upon activation. However, due to the permanent deformation of the wires, the force delivery of beta-titanium 0.019" x 0.025" wires decreased more rapidly than the other three types of rectangular wires. Both TorQ A and B wires were exerting more force than Nitinol Classic 0.019" x 0.025" wires prior to TorQ A and B wires cracking or crazing.

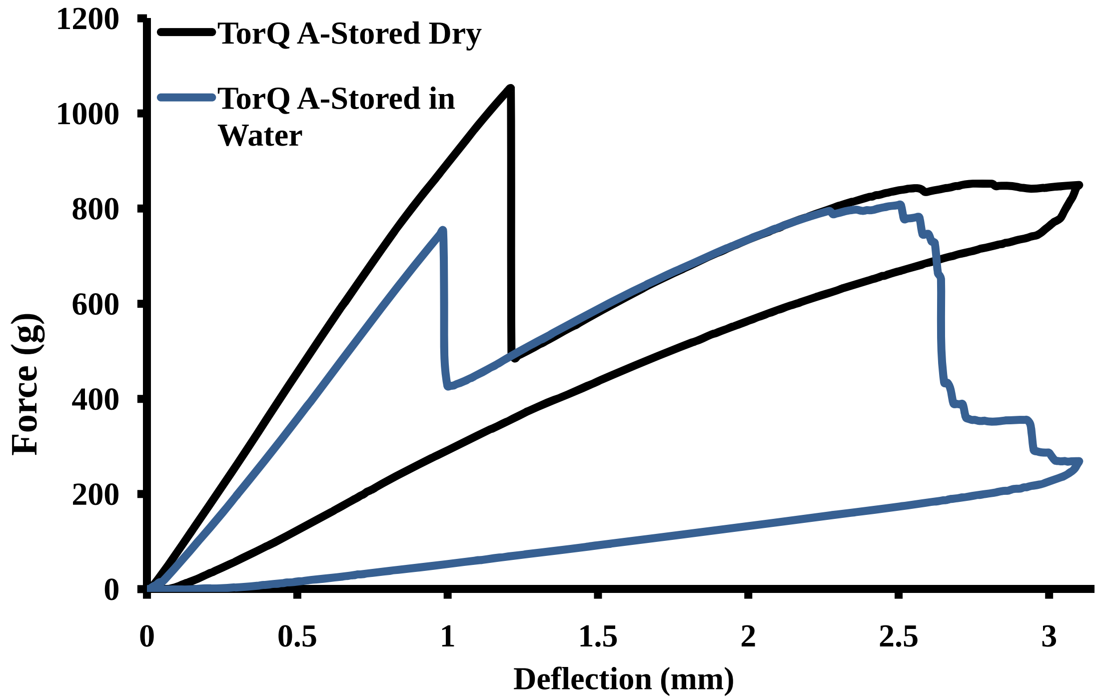

Figure 17. Graph example of force-deflection curve for TorQ A.

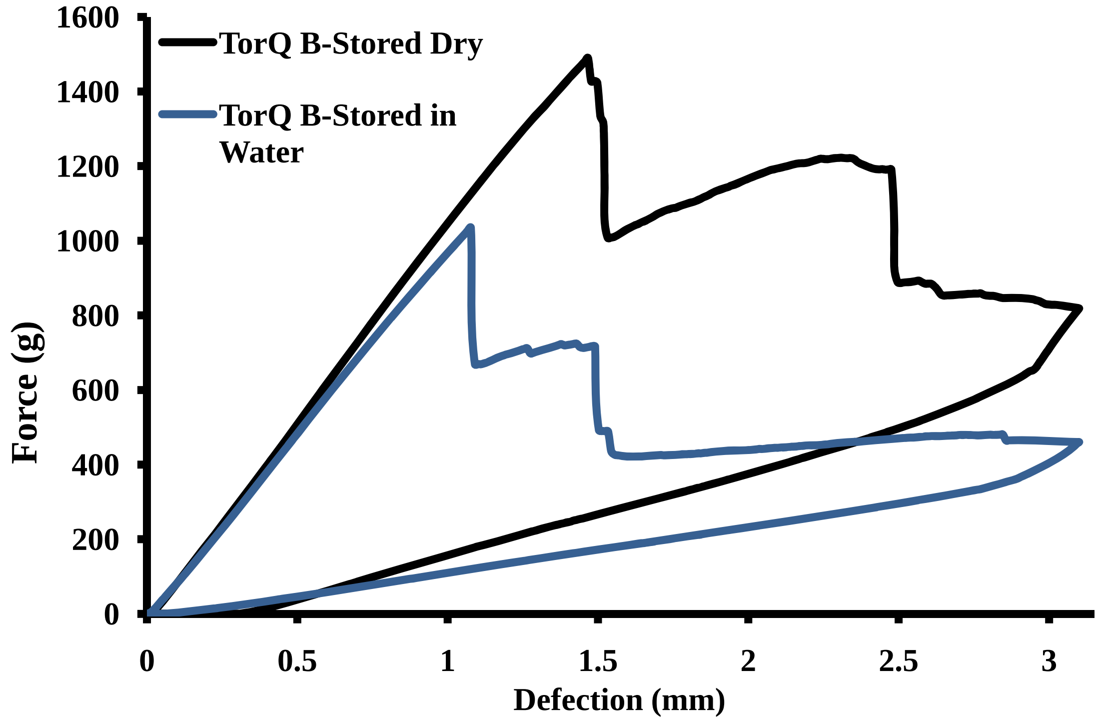

Figure 18. Graph example of force-deflection curve for TorQ B.

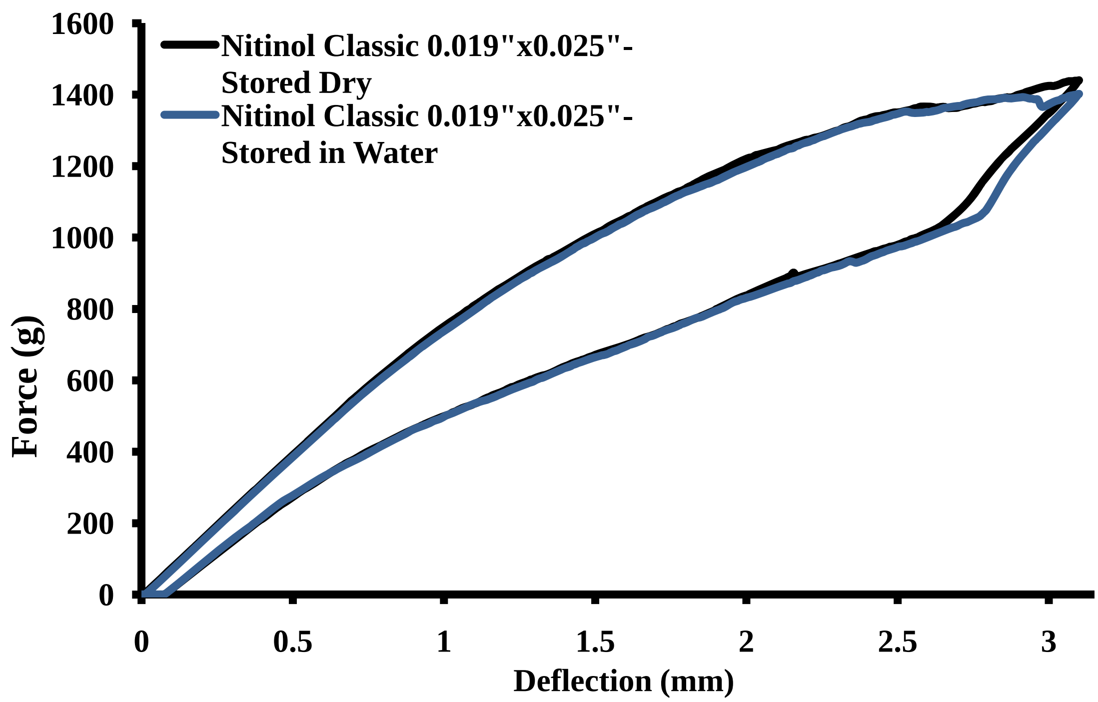

Figure 19. Graph example of force-deflection curve for Nitinol Classic 0.019” x 0.025”.

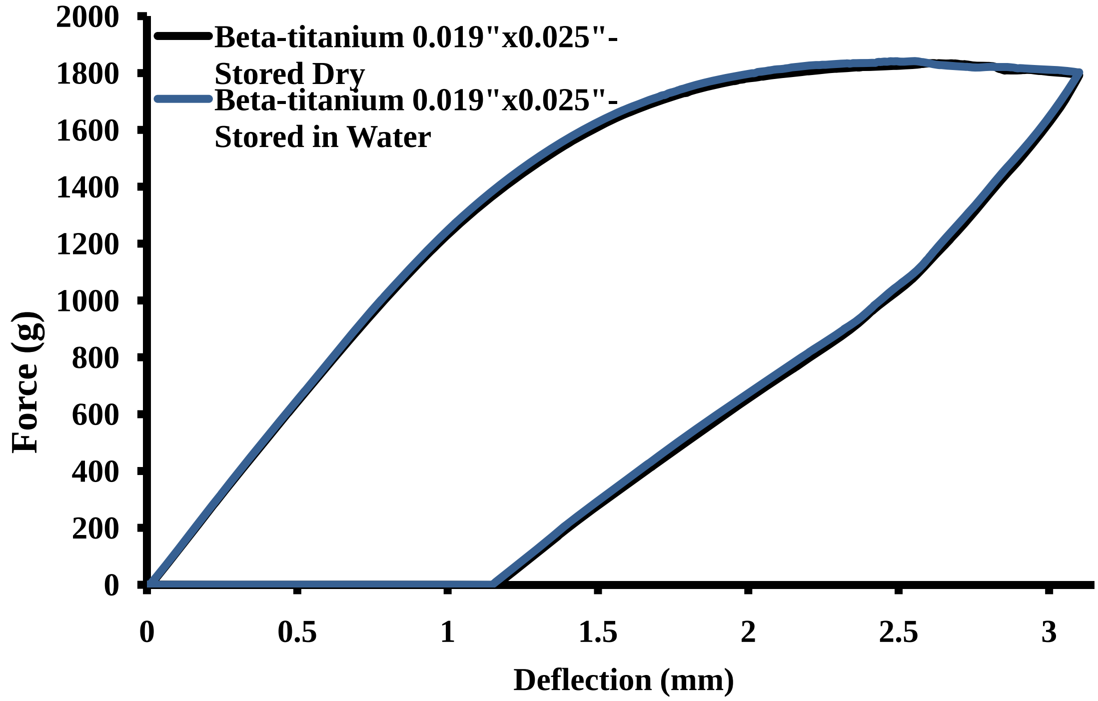

Figure 20. Graph example of force-deflection curve for beta-titanium 0.019” x 0.025”.

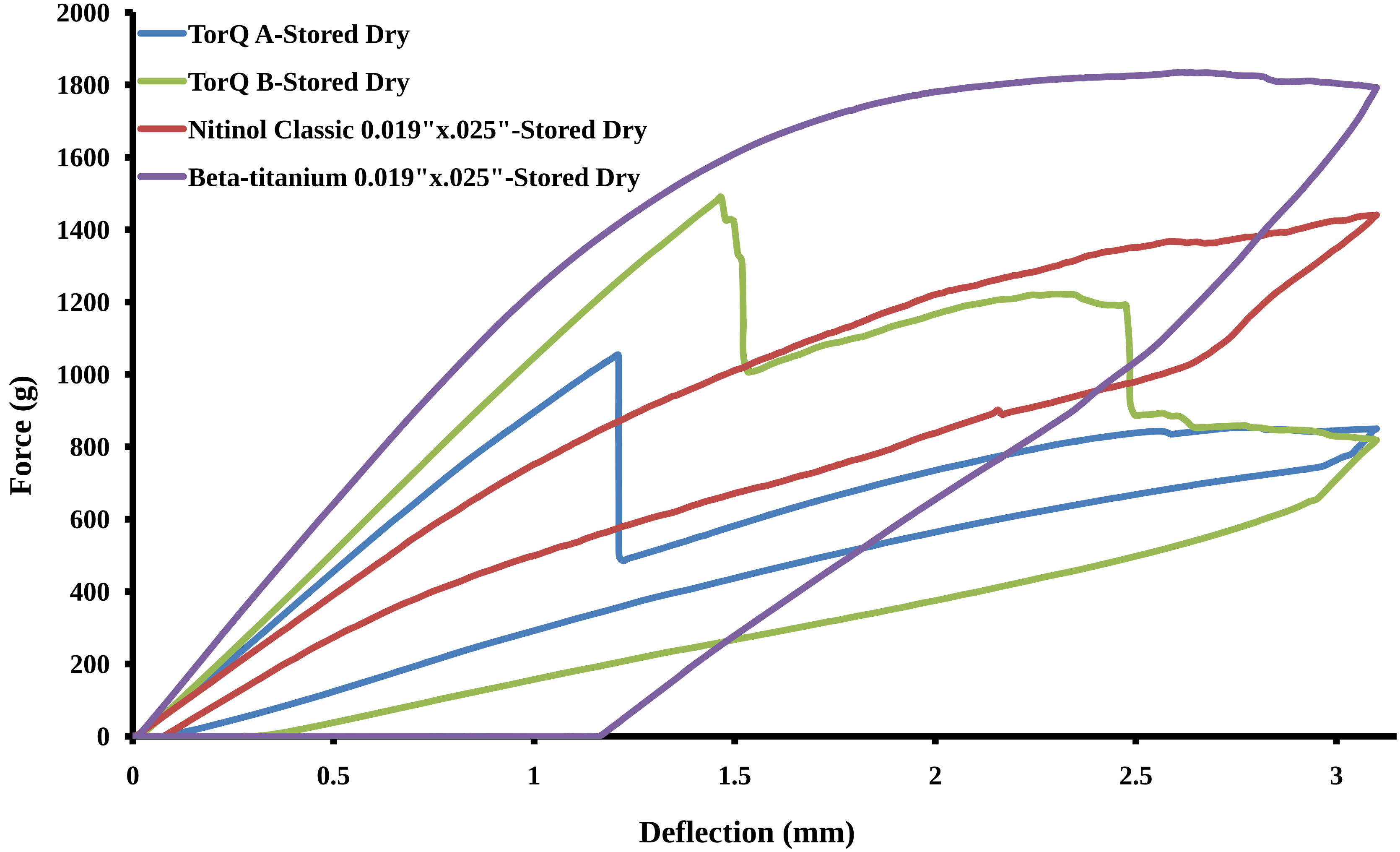

Figure 21. Comparison of force-deflection curves of rectangular wires in the stored dry groups.

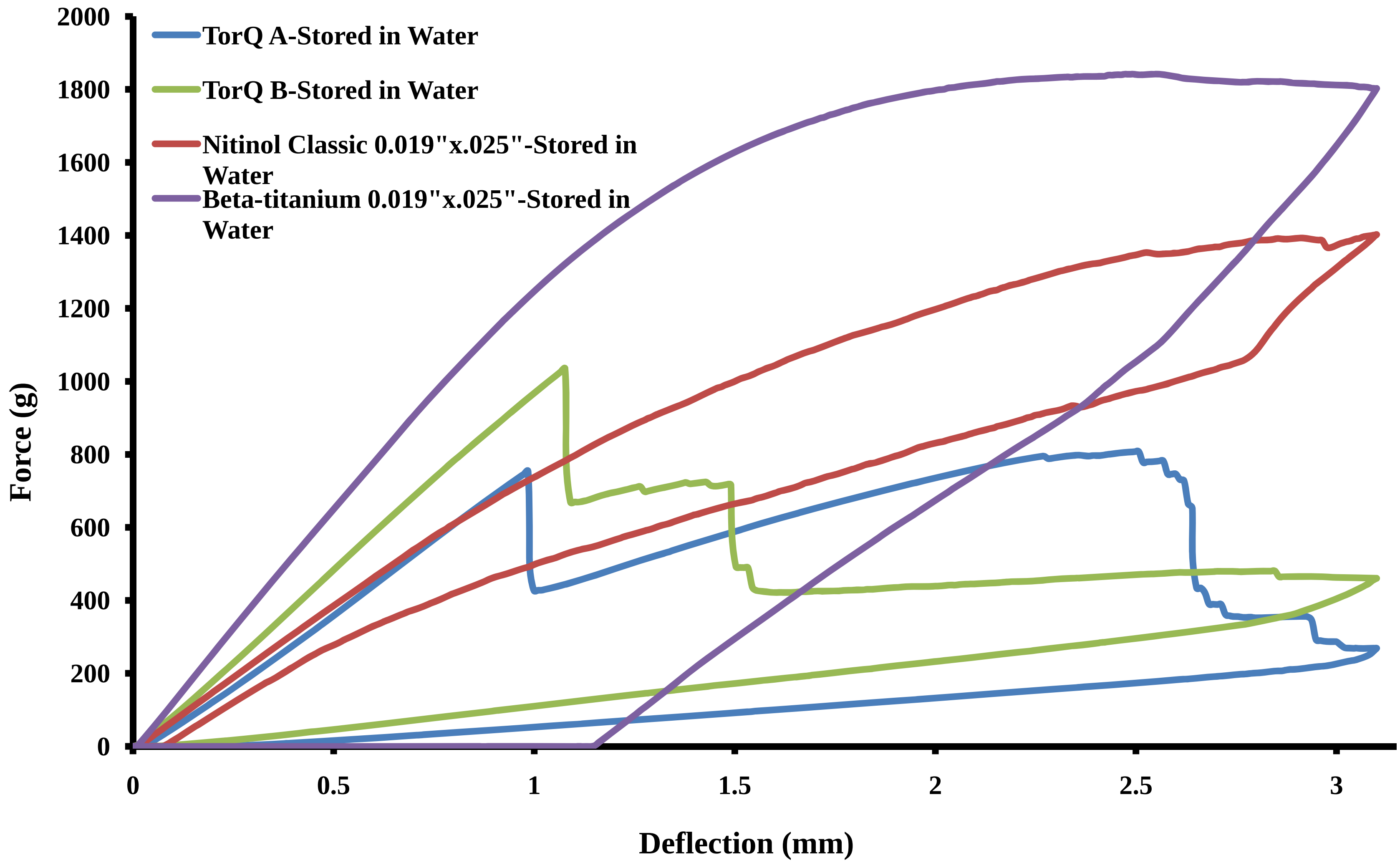

Figure 22. Comparison of force-deflection curves of rectangular wires in the stored in water groups.

The bending values during activation and deactivations for rectangular wires are presented in Tables 5 and 6. During activation for rectangular wires, beta-titanium had the highest stiffness and bending modulus. Interestingly, these two values were significantly different between the stored dry and stored in water groups. All the TorQ A and TorQ B wires cracked during the bending test. Despite cracking, the stored dry groups of the fiber-reinforced composite wires still delivered higher force levels compared to their corresponding stored in water groups. Nitinol Classic 0.019" x 0.025" wires had the tendency to flip from edgewise to flat-wise. As soon as the wires flipped, the bending test was stopped. Therefore, the force level at 3 mm was not applicable.

For deactivation of rectangular wires (Table 6), beta-titanium wires still had the highest stiffness and modulus. Since they were significantly permanently deformed, they delivered no force at 1 mm of deflection. The TorQ A stored dry group was significantly different from the stored in water group for all values in the deactivation table. The same applies to TorQ B, except the elastic recovery values.

Table 5. Bending values during activation for rectangular wires.

ARCHWIRE	ACTIVATION					
	Stiffness (g/mm)	Modulus (GPa)	Force at 1 mm (g)	Force at 2 mm (g)	Force at 3 mm (g)	% with cracks (at deflection)
TorQ A (0.019" x 0.025")-Dry	857 ± 215 C	28.0 ± 6.4 C	771 ± 224 C	767 ± 27 D*	786 ± 246 B*	100 (1.12 ± 0.23 mm)
TorQ A (0.019" x 0.025")-Water 30 d,37^{o}C	843 ± 73 C	30.4 ± 1.3 C	744 ± 162 C	572 ± 187 D*	360 ± 198 B*	100 (1.10 ± 0.20 mm)
TorQ B (0.021" x 0.025")-Dry	1162 ± 114 B	28.9 ± 2.3 C	973 ± 257 B*	1005 ± 107 C*	656 ± 329 B*	100 (1.17 ± 0.22 mm)
TorQ B (0.021" x 0.025")-Water 30 d,37^{o}C	1100 ± 138 B	27.8 ± 5.2 C	819 ± 253 B*	732 ± 296 C*	350 ± 145 B*	100 (0.99 ± 0.13 mm)
Beta-titanium (0.019" x 0.025")-Dry	1274 ± 15 A*	72.9 ± 0.9 A*	1244 ± 44 A	1801 ± 25 A	1813 ± 24 A	0
Beta-titanium (0.019" x 0.025")-Water 30 d,37^{o}C	1285 ± 14 A*	73.5 ± 0.8 A*	1252 ± 15 A	1813 ± 22 A	1824 ± 35 A	0
Nitinol Classic (0.019" x 0.025")-Dry	765 ± 11 D	44.8 ± 0.7 B	739 ± 13 C	1184 ± 31 B	N/A	0
Nitinol Classic (0.019" x 0.025")-Water 30 d,37^{o}C	762 ± 10 D	44.7 ± 0.6 B	738 ± 8 C	1188 ± 20 B	N/A	0

Within each parameter, different letters denote significant differences ($p<0.05$) exist between types of wire.
* indicates a significant difference between dry and water-stored wires of the same type/size.
Note: Nitinol Classic wires tended to flip to a flat-wise orientation during bending above 2 mm deflection. Data not presented for subsequent deflections.

Table 6. Bending values during deactivation for rectangular wires.

ARCHWIRE	DEACTIVATION					
	Stiffness (g/mm)	Modulus (GPa)	Force at 3 mm (g)	Force at 2 mm (g)	Force at 1 mm (g)	Elastic Recovery (%)
TorQ A (0.019" x 0.025")-Dry	254 ± 119 B*	8.3 ± 3.7 B*	641 ± 255 B*	448 ± 202 B*	229 ± 109 A*	97.1 ± 3.3 A*
TorQ A (0.019" x 0.025")-Water 30 d,37^{o}C	59 ± 41 B*	2.2 ± 1.5 B*	253 ± 121 B*	120 ± 74 B*	51 ± 35 A*	91.1 ± 5.9 A*
TorQ B (0.021" x 0.025")-Dry	161 ± 104 C*	3.9 ± 2.5 C*	550 ± 290 B*	304 ± 187 B*	139 ± 93 B*	94.6 ± 3.5 A
TorQ B (0.021" x 0.025")-Water 30 d,37^{o}C	65 ± 47 C*	1.7 ± 1.3 C*	270 ± 129 B*	138 ± 78 B*	56 ± 39 B*	90.1 ± 5.5 A
Beta-titanium (0.019" x 0.025")-Dry	753 ± 15 A	176.2 ± 3.4 A	1617 ± 13 A	636 ± 16 A	0	62.3 ± 0.5 B
Beta-titanium (0.019" x 0.025")-Water 30 d,37^{o}C	756 ± 16 A	176.7 ± 3.8 A	1628 ± 20 A	643 ± 13 A	0	62.5 ± 0.6 B

Within each parameter, different letters denote significant differences ($p<0.05$) exist between types of wire.
* indicates a significant difference between dry and water-stored wires of the same type/size.
Note: Nitinol Classic 0.019" x 0.025" wires tended to flip to a flat-wise orientation during bending above 2 mm deflection. Data not presented.

CHAPTER 5
DISCUSSION

Composites reinforced with long, continuous fibers have been used in many areas. They are useful in aerospace, automotive, and sports because of their high strength to weight ratio (Valiathan & Dhar, 2006). For orthodontics, the value of fiber-reinforced composite mostly lies in its translucent optical property. The combination of polycarbonate or ceramic brackets with fiber-reinforced composite archwires provides the best esthetic appliance option for buccal fixed orthodontics. Other advantages of fiber-reinforced composite include the ability to make wires of the same cross-section with different stiffness values by varying the fiber and resin ratio. Instead of soldering and welding, attachments can be directly bonded onto the fiber reinforced archwires. Composite wires are possible alternatives to alloy wires when there is a concern for nickel allergy. Being metal-free, these archwires can be left in place for nuclear magnetic resonance imaging (Valiathan & Dhar, 2006). Although fiber-reinforced composite archwires have many advantages, they should also possess clinically desirable mechanical properties compared to the existing alloy wires.

In this study, the dimensions of all wire segments were measured. The average dimensions of all wires were different from the dimensions specified by manufacturers. All of the round wires and alloy rectangular wires were measured to be within 5% of that stated by the manufacturers. However, for the rectangular fiber-reinforced composite wires, the dimensions varied from expected by 7 to 21%. Overall, the dimensions of the alloy wires were consistent among the same group and along a segment, but the dimensions of the fiber-reinforced composite wires were not. This inconsistent variation from the specified dimension could cause the composite archwires to not fit in the slot of the brackets. If they fit in the slots, there might be an increase in friction if the sizes are greater than expected. Therefore, utilization of composite wires in space closure using sliding mechanics might result in uneven space closure or reduced efficiency. With the variation in dimension, the force values are then different from expected and are unpredictable.

Composite and alloy wires were tested using 3-point bending. The rectangular wires had larger dimensions than the round wires that were tested; therefore as expected, the rectangular wires had greater stiffness than round wires. The stiffness values in descending order were 0.019” x 0.025 beta-titanium, TorQ B (0.021” x 0.025”), TorQ A (0.019” x 0.025”), 0.019” x 0.025” Nitinol Classic, 0.016” stainless steel, Align C (0.021”), 0.018” Nitiniol Classic, Align B (0.019”), 0.016” Nitinol Classic, Align A (0.018”), and 0.014” Nitinol Classic. These stiffness values were all significantly different from each other. Rectangular fiber-reinforced composite wires had smaller stiffness comparing to beta-titanium of the same size, but slightly higher stiffness than Nitinol Classic of the same size. For round wires, composite wires had a lower stiffness than stainless steel and Nitinol Classic archwires of the same size. The force delivery values corresponded with the stiffness values well, until cracking and cracking occurred in the fiber-reinforced composite wires. Since the rectangular composite wires had larger dimensions comparing to the round wires, they cracked/crazed with smaller deflections. Due to cracking, although TorQ A and B had significantly higher stiffness than 0.019” x 0.025” Nitinol Classic, there was no significant difference between the force delivery of TorQ A and 0.019” x 0.025” Nitinol Classic at 1 mm of deflection; but at 2 mm of deflection, 0.019” x 0.025” Nitinol Classic had a significantly greater force value than TorQ A and B. The effect of cracking among the round wires started to show at deflections of 3 mm where force delivery levels of Align A, B, and C were significantly lower than 0.018” Nitinol Classic. It is important clinically to know that composite wires are not as stiff as the stainless steel and beta-titanium wires of the same size because this makes them less suitable for certain types of mechanics that requires rigid archwires, like closing spaces using sliding mechanics, correcting anterioposterior relationships using inter-arch elastics, or maintaining transverse dimension. For the round wires, fiber-reinforced composite archwires had less stiffness than Nitinol Classic of the same size. This means the composite round wires could fill up the bracket slot more while delivering more gentle forces than Nitinol Classic.

The apparent discrepancy in comparison between the rectangular and round wires with respect to composite versus Nitinol Classic may be related to the actual size of the wires when measured instead of relying on the manufacturer-specified dimensions. For instance, the rectangular NiTi wires were less stiff than the “same size” TorQ A, but its modulus was larger when computed out with the actual dimensions factored. This is in contrast to the round wires where the “same size” NiTi was stiffer and had a greater modulus. Due to the dimensions being closer to stated and less differential between NiTi, the stiffness and modulus followed the same trend.

No significant difference was detected in the stiffness of the alloy wires between two storing methods, except 0.019” x 0.025” beta-titanium. The stored in water group of 0.019” x 0.025” beta-titanium wire had statistically significantly higher stiffness than the stored dry group by 11 g/mm. Since this difference in stiffness was very small, which was only less than 1% of the total stiffness, and the force delivery levels for the two groups were not significantly different, this difference in stiffness is clinically insignificant. The adverse effect on fiber-reinforced composite wires was demonstrated in the percentage of wires cracked. For round wires, the stored in water groups had a higher rate of crack occurrences compared to their corresponding store dry groups. For Align C, the force level at 3 mm of deflection in the stored in water group was significantly lower than the stored dry group. Although the cracking rates were 100% for the rectangular fiber-reinforced composite groups, the stored in water group for TorQ A and TorQ B had significantly lower force delivery levels starting at 1 mm and 2 mm deflections, respectively. The force level at any given deflection for alloy archwires was not significantly different between stored in water and stored dry groups. Alloy archwires were not significantly affected by water because water cannot diffuse into alloys. Although surface corrosion is possible, a period of 30 days is too short for it to cause an effect when stored in only water. For fiber-reinforced composite, no corrosion will occur, but water can diffuse into the resin matrix and act as a plasticizer, defined as a "material incorporated into a plastic to increase its workability and flexibility or distensibility" (Pebly, 1987). Water molecules make the movement of polymer chains easier under stress (Chai et al., 2005). At the molecular level, the absorbed water molecules can disperse in the resin matrix randomly or interact with specific sites of the resin backbone. In either case, water absorption results in a decreased glass transition temperature and declined mechanical properties of the fiber-reinforced composite. With simple diffusion, the glass transition temperature is reduced due to a complex relationship associated with the increased volume. With specific interaction with the backbone, the decrease in glass transition temperature and mechanical properties is due to redistribution of molecular bonds and formation of hydrogen bonds (Cotugno et al., 2002). Hydrolytic degradation of resin corresponds to the lower force level delivery of the wires in the stored in water group, however, it does not explain why these wires had a higher cracking rate. Therefore, there is possibly another mechanism at play. One possible mechanism is the glass fibers in the composite wires also experienced hydrolytic degradation which makes them break easier or possibly the bond between the fiber and resin matrix was compromised, leading to alterations in stress transfer.

These findings are important as a clinical guideline for using fiber-reinforced composite wires from BioMers. According to a force level comparison given by BioMers (Table 7), Align A, B, C, and TorQ A, B should have similar forces value as 0.016” NiTi, 0.018” NiTi, 0.016” SS, 0.019” x 0.025” NiTi, and 0.019” x 0.025” beta-titanium, respectively. All fiber-reinforced composite wires had lower force delivery level than BioMers specified, except for TorQ A. Although the force levels were not the same, they were fairly comparable, except for Align C. The force level of Align C would probably be more comparable to that of 0.020” Nitinol Classic, instead of 0.016” stainless steel.

Table 7. Force levels and deflection limits specified by BioMers.
(Adopted from http://www.simpliclearpro.com/resource-library/resources/)

Archwire	Dimension	Forces similar to	Deflection limit
Align A	0.018”	0.016” NiTi	2-3 mm
Align B	0.019”	0.018” NiTi	1-1.5 mm
Align C	0.021”	0.016” SS	0.5 mm
TorQ A	0.019” x 0.025”	0.019” x 0.025” NiTi	0.5 mm
TorQ B	0.021” x 0.025”	0.019” x 0.025” beta-titanium	0.5 mm

It is also clinically important to know the deflection limit to prevent damage or permanent deformation to the wires. BioMers also provided deflection limit guidelines for its fiber-reinforced composite wires (Table 7); however, the company did not specify the length of the span for these deflections. That piece of information is important because the stress on the wire with a 3-mm deflection over a 14-mm span is very different from that with a 3-mm deflection over a 5-mm span. In this study, all deflections were over a 14-mm span which is the average distance between the labial center of a mandibular lateral incisor and a first premolar on the same side of the arch (Nakano et al., 1999; Cacciafesta et al., 2008). Under the stored dry condition, Align A could be deflected up to 3 mm; the cracking of Align B and C happened around 2.5 mm of deflection; TorQ A and B cracked around 1 mm. Under the store in water condition, 30% of Align A wire segments cracked around 1.4 mm which is much lower than the specified deflection guide for Align A. Therefore, to prevent wire damage, Align A probably should not be deflected more than 1 mm clinically. This means although the force level for Align A and 0.016” Nitinol Classic are similar, they can’t be utilized the same way clinically. For the stored in water segments of Align B and C, the cracking rates increased comparing to the stored dry groups, but the average deflection limits before cracking were not significantly different. For TorQ A and B, the crack rates were the same in both storage groups. Although the TorQ A and B wires stored in water group cracked with less deflection, it was not statistically or clinically significant. Therefore, to utilize fiber-reinforced composite archwires from BioMers successfully, it is critical to keep their suggested deflection limits in mind with the modification of deflection limit for Align A to be 1 mm. If the wires crack, they still exert some forces, but they are much less than without cracks. This decreased force level after cracking is even more severe with water immersion.

This study helps clinicians understand some mechanical properties of fiber-reinforced composite wires by comparing them to alloy archwires that are already familiar to orthodontists. There are some limitations in this study. First, since the rectangular wires were tested edgewise, which corresponds to in-and-out and rotation movements, it is also important to find out if the results would be better if the wires were bent flat-wise, for up-and-down and tipping movements, in a future study. Second, this study used the deflection of 3.1 mm as a guideline from International Organization of Standardization, but that extension is not practical for some wires. Therefore, future studies should use more practical deflection tailoring to the archwires. Also instead of loading the archwires only once to a 3 mm deflection, loading the wires multiple times with smaller deflections better simulates the oral environment during mastication. Third, although thirty days is usually the average interval between conventional orthodontic adjustment appointments, with the advancement of orthodontic appliances, for example, nickel titanium close coil springs, superelastic nickel titanium archwires, and self-ligating brackets, many orthodontists sometimes extend their adjustment interval to six to eight weeks. Therefore, it is also important to study the effect of water on fiber-reinforced composite wires with longer immersion time. The fourth limitation of this study is using distilled water for storage instead of using saliva. Saliva is composed of 99% water and various electrolytes: sodium, potassium, calcium, chloride, magnesium, bicarbonate, and phosphates. Saliva also contains proteins, enzymes, immunoglobulins and other antimicrobial factors, mucins, albumin, polypeptides, oligopeptides, and nitrogenous products, like urea and ammonia (Humphrey & Williamson, 2001; de Almeida et al., 2008). Normal pH for saliva is between 6 and 7; therefore it is slightly acidic. With different levels of salivary flow rate, the pH of saliva can range from 5.3 with low flow to 7.8 with peak flow (de Almeida et al., 2008). It is important to know how fiber-reinforced composite archwires perform after prolonged enzymatic and acidic challenges in addition to hydrolytic degradation. The fifth limitation is the three-point bending test. Three-point bending is great in testing stiffness and bending modulus of the wires, however, the archwires are used clinically along with brackets and ligatures, creating numerous points of contacts and more friction for sliding. Therefore, future studies could use three brackets for the test to simulate the clinical environment or conduct an in vivo study.

CHAPTER 6
CONCLUSION

In this study, the following were demonstrated:

- The force levels and deflection limits for fiber-reinforced composite suggested by BioMers are comparable to the current finding, except:
    - The force level of Align C is much lower than that of 0.016” stainless steel. Therefore, the force level of Align C is more comparable to that of 0.020” Nitinol Classic.
    - The suggested deflection limit for Align A after water storage should be 1 mm instead of 2-3 mm.
- Water immersion for thirty days was damaging to the fiber-reinforced composite archwires. The larger composite wires were affected more, as they were more likely to crack/craze during bending, resulting in decreased amounts of force applied at a given deflection.
- Alloy wires were not significantly affected by water storage.
- Overall, the alloy wires possessed vastly more consistent force values compared to the composite wires.
- Although the translucent archwires from BioMers present a more esthetic option for patients, their mechanical response is less reliable than alloy wires, possibly compromising treatment efficiency.

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