Orthodontic Open-Coil Spring Deactivation Forces Differ with Varying Activation Levels

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ORTHODONTIC OPEN-COIL SPRING DEACTIVATION FORCES DIFFER WITH VARYING ACTIVATION LEVELS

by

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ABSTRACT

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Objectives:
Coil springs are used in orthodontics to deliver forces to move teeth. However, the optimal amount of activation to produce predictable, continuous deactivation forces, for the purpose of orthodontic tooth movement, is unclear. The purpose of this study was to evaluate the deactivation force characteristics of nickel-titanium open-coil springs after varying levels of activation.

Methods:
Four open-coil spring products were evaluated: Dentsply GAC 100 gram and 150 gram, Orthoclassic Orthodontics Medium 150 gram, and American Orthodontics nickel-titanium springs. Open-coil springs were compressed (activation) with a universal testing machine to 20, 40, 60, or 80% of original length (15 mm; n=15/product/activation level). Measurements were conducted at 37^{o}C to simulate intraoral temperatures. Deactivation force values at 3 mm were compared among the four activation levels within each product using ANOVA/Tukey.

Results:
For all four open-coil spring products, deactivation force values at 3 mm significantly ($p<0.05$) decreased with greater activation/compression level. When activated to 20% of original length, the force values at 3 mm compression upon deactivation were only 53-73% of the force level when activated to 80% of original length. When activated to 40% of original length, the force values at 3 mm compression upon deactivation were 65-91% of the force level when activated to 80% of original length.

Conclusions:
Deactivation forces of nickel-titanium open-coil springs at a given compression level are dependent upon the extent of prior compression. Orthodontists should be aware that force values in open-coil springs depend not only on material and structural factors, but amount of activation as well.

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CHAPTER 1
INTRODUCTION

Orthodontics is the specialized field of dentistry devoted to "the study of the growth of the craniofacial complex, the development of occlusion, and the treatment of dentofacial abnormalities" (Moyers, 1988). The word "orthodontics" is derived from the Greek *orthos*, meaning straight, and *odont*, meaning tooth (Moyers, 1988). It is estimated that 57 to 59% of the majority of all racial and ethnic groups in the United States has an orthodontic treatment need (Proffit, Fields, & Moray, 1998). The straightening of misaligned teeth is critical to the practice of orthodontics and is a vital component of the treatment of dentofacial abnormalities. Due to the complex nature of the structural and functional components that support the teeth in the jaws, called the periodontal apparatus, teeth can be moved within their bony housing via the application of a force. These tooth-moving forces can be generated by many entities including, but not limited to, the pressure of the cheeks and tongue, mastication, extra-oral appliances, and intra-oral appliances utilizing brackets, wires, screws, elastomerics and coil springs. Each of these entities is utilized in a different manner to produce orthodontic tooth movement, and it is important for the orthodontist to appreciate the force characteristics produced by each.

Coil springs are an integral part of the orthodontic intra-oral appliance. They are manufactured primarily in open-coil and closed-coil forms. Generally, open-coil springs are used to make space between teeth and closed-coil springs are used to close space between teeth.

When a force is applied to a tooth for a sustained period of time, the tooth will tend to shift position. This phenomenon can be explained by the pressure-tension theory of tooth movement (Proffit, Fields, & Sarver, 2013). The theory explains that when a force is applied to a tooth (directly at the center of resistance), an area of pressure is created on the side of the periodontal ligament opposite the point of force application, and an area of tension is created on the side of the periodontal ligament immediately adjacent to the point of force application. The area of pressure in the periodontal ligament causes a cascade of cellular and chemical events leading to adjacent bone resorption, and the area of tension leads to another cascade of cellular and chemical events, but this time, leading to bone deposition. The end-result is tooth movement. There are two types of resorption that can take place: frontal and undermining. Frontal resorption leads to more efficient and less painful tooth movement, whereas undermining resorption leads to the opposite (Proffit, Fields, & Sarver, 2013). Frontal resorption is usually created with relatively lighter forces and undermining resorption with relatively heavier forces. The amount of force necessary to cause movement is dependent on the surface area of the periodontal ligament, which differs for each tooth. Bodily movement of teeth with primarily frontal resorption generally requires 70-120 grams of force (Proffit, Fields, & Sarver, 2013). Exceeding this amount can cause detrimental clinical effects, like pain, loss of rotational control and anchorage loss (Yee, Turk, Elekdag-Turk, Cheng, & Darendeliler, 2009).

Not only is force magnitude important to efficient and less painful tooth movement but force duration as well. Proffit et al. (2013) suggests the force duration threshold required for orthodontic tooth movement in humans is four to eight hours. Furthermore, Proffit et al. (2013) suggests tooth movement efficiency is a function of force duration, with a constant force producing near maximal efficiency. Put together, relatively light forces, in the range that produce frontal resorption, and constant forces will produce the most efficient tooth movement. In other words, continuous light force should be used in an ideal orthodontic appliance (Mirhashemi, Saffarshahroudi, Sodagar, & Atai, 2012).

In order for an activated coil spring to produce light, near constant deactivation forces, a number of factors need to be considered in their manufacture. These factors include pitch, diameter of the wire and the coils, among others, but perhaps the most important factor is the material. Nickel-titanium alloy has long been utilized in the orthodontic intraoral appliance due to its superelastic behavior (Miura, 1986). When some nickel-titanium wires are bent or stressed, the load-deflection curve will sometimes show an area with a slope close to zero, which means as the wire straightens, a relatively constant force is produced. Nickel-titanium wires can also exhibit a relatively long range of activation before permanent deformation takes place, which means these wires can be bent and distorted to a relatively high degree and return to their original shape (Burstone, Bai, & Morton, 1985). The superelastic behavior of nickel-titanium makes it ideally suited as the material for coil springs and other components of the intraoral appliance.

Elastomerics are frequently used in orthodontics for similar purposes as coil springs, however, they pose a significant disadvantage: elastomerics do not exhibit superelastic behavior. Rapid force decay is a significant problem as well (Halimi, Benyahia, Doikkali, Azeroual, & Zaoui, 2012). Elastomeric chain force degradation has been reported to be as high as 59-69% in the first hour (Weissheimer, Locks, de Menezes, Borgatto, & Derech, 2013). The rapid force decay affects mechanical properties of the elastomeric chains as well as their clinical efficacy (Halimi, Benyahia, Doikkali, Azeroual, & Zaoui, 2012). Overall, depending on the clinical application, NiTi springs can be more adequate for orthodontic tooth movement (Santos, Tortamano, Naccarato, Dominguez-Rodriguez , & Vigorito, 2007). This illustrates the importance of coil springs in the orthodontist's armamentarium.

Once an orthodontist has chosen to use a nickel-titanium open-coil spring, for example, to open space for a tooth that is blocked out of the arch, he or she must decide how much force is required to accomplish the tooth movement efficiently. Most manufactures and orthodontic supply companies will list the amount of force produced by a particular nickel-titanium open-coil spring. Those reported force levels may or may not be accurate in clinical conditions. Another decision the orthodontist must make is how much to compress the coil. A coil compressed to 80% of the original length may produce a different amount of force than the same length of coil compressed to 20% of the original length. In other words, the superelastic behavior may not be observed at all levels of compression, and different force levels could be produced at varying amounts of compression. Overall, it is important for the orthodontist to have confidence in knowing how much force the appliance is producing.

Being aware of the biomechanics and force characteristics of the intraoral appliance is an important part of orthodontic knowledge and training. As a commonly used instrument in the orthodontist's armamentarium, the open-coil spring should be well understood. Material and structural factors of the open-coil spring play a role in force production, but the amount of activation can affect deactivation forces as well. When coils are used and activated in the intraoral appliance, only the rebound or deactivation force is used to move teeth. It is hypothesized that as the amount of initial activation of a nickel-titanium open-coil spring increases, the resultant deactivation force will also increase.

CHAPTER 2
LITERATURE REVIEW

Coil Springs Emerge In Biomaterials Research

Open-coil springs have been the topic of research since the 1930s. Early studies focused on the comparison of open-coil springs made of precious and non-precious metals. Non-precious metals, such as steel, were found to produce more force at a given activation than precious metals, such as gold or silver (Arnold & Cunningham, 1934), (Johnson, 1934). Johnson followed this initial research with an analysis of the effect of spring wire diameter, arch wire type, spring length, lumen size, and again, the effect of precious versus non-precious metals. In the 1950s, Born and Bell continued this theme and published data illustrating the effect of wire size, arch wire size, and lumen size, at varying degrees of compression, on deactivation force production (Bell, 1951) (Born, 1955). These early studies, among others, helped to shed light on open-coil spring deactivation forces; however, the studies were limited because the reported deactivation forces could only be recorded at a discrete level of activation. Later studies utilized the continuous load-deflection curve to illuminate the broader picture of open-coil spring behavior.

Chaconas et al. (1984) produced one of the first data sets, from the compression of open-coil springs, with load-deflection curves. The study examined three different materials: stainless steel, chrome steel alloy, and Elgiloy alloy. Lumen size and coil wire thickness were examined as well. Lastly, the arch wire (the wire that is threaded into the coil spring and used in the testing apparatus) was varied between rectangular and round. The coil springs were 20 mm in length and were compressed 10 mm (1/2 the original length). An Instron test machine was used to record load deflection data. Chaconas et al. found that as lumen size increased, deactivation forces decreased, and as coil wire size increased, deactivation forces increased. Varying arch wire shape only produced a significant difference in coil springs with larger lumen sizes. Coil springs with smaller lumen sizes were not significantly affected. Lastly, there were differences between manufacturers of coil springs. The load deflection curves varied for coil springs reportedly of the same size and material, but produced by different manufacturers. This was attributed to differences in pitch, or the number of loops in a given section of coil. Those manufacturers that incorporated more total wire in a given length of coil, meaning a higher pitch or more loops, produced lighter forces overall. All of the load deflection curves were reportedly similar in shape. There was an initial, relatively short period of non-linearity, attributed to the spring "settling in" to the test apparatus. Immediately thereafter, the curve was linear, meaning as the amount of compression increased, the amount of force also increased in a relatively proportional fashion. Chaconas et al. recommends that orthodontists compress an open-coil spring, of one of the aforementioned materials, to 1/3rd its original length for clinical applications. This should produce force levels capable of tooth movement. Compressing springs greater than this amount could produce orthopedic forces, which are too great for efficient tooth movement (Chaconas, Caputo, & Harvey, 1984).

Nickel-Titanium Alloy

Nickel-titanium alloy (NiTi) was first developed by the US Naval Ordnance Laboratory in the 1960s and was named *nitinol* by combining the first letters of nickel, followed by titanium, and lastly Naval Ordnance Laboratory (Proffit, Fields, & Sarver, 2013). NiTi alloy is especially useful in orthodontics due to its dual martensite and austenite crystal structure, and, more specifically, its fully reversible transition between the two. This transition can occur at relatively low temperature (Proffit, Fields, & Sarver, 2013). NiTi wires are able to withstand great amounts of stress before permanent deformation occurs relative to other metals and alloys, for example, stainless steel, and NiTi wires have a low load deflection rate. As a NiTi wire deactivates, or returns to its original shape, the resultant deactivation forces are relatively constant and light. This behavior is largely due to the phase transformation from martensite to austenite (Proffit, Fields, & Sarver, 2013).

Improvements in NiTi alloy were developed in the 1980s. These improvements lead to a phenomenon known as "super-elasticity". The original *nitinol* wire was found to have good springback, but lacked this super-elasticity (Miura, 1986). A super-elastic wire is one that, after being bent or activated to a point before permanent deformation, displays a near constant amount of deactivation force as the wire returns to its original, unbent shape (Miura, 1986). Miura and colleagues reported on a new type of NiTi wire developed by the Furukawa Electric Co., LTD of Japan (Miura, 1986). This wire was shown to have this super-elasticity phenomenon. Chinese NiTi, introduced around that time by the General Research Institute for Non-Ferrous Metals in Beijing, China, was found to have a lower load deflection rate, higher springback, and more constant deactivation forces compared to *nitinol* wires (Burstone, Bai, & Morton, 1985).

Nickel-Titanium Properties

"Superelasticity" is a property of NiTi determined by the tri-dimensional lattice of the alloy, which can be present in the martensite and austenite phases (Santoro, Nicolay, & Cangialosi, 2001). The martensitic phase is characterized by a lattice that is body-centered (cubic or tetragonal), and the austenitic phase is characterized by a lattice that is face-centered (hexagonal close packed). The concentrations of these two phases can be modified by temperature variation, such that the crystal structure undergoes molecular arrangement modification without atomic composition changes. In other words, the austenite to martensite transformation is reversible. At lower temperatures, the NiTi alloy is primarily in the martensitic phase, and at higher temperatures, the alloy is primarily in the austenitic phase. The temperature at which the transformation from martensite to austenite occurs, and visa versa, is called the temperature transitional range (TTR). The TTR varies based on the unique composition and material processing history of each NiTi alloy (Santoro, Nicolay, & Cangialosi, 2001).

Evidence of superelasticity can be seen on a stress-strain or load-deflection graph. As a superelastic NiTi wire is initially stressed or deflected, the corresponding amount of strain or load increases in a relatively linear or proportional pattern. This occurs within the austenite phase. Once a critical stress is reached, further strain results in transformation to martensite and the slope of the graph becomes flatter. As the stress or deflection is decreased, the corresponding amount of strain or load does not follow a linear or proportional pattern. Rather, there is a "superelastic plateau" where the strain or load remains constant. If this superelastic plateau is absent, then the NiTi wire is not exhibiting superelastic behavior. If a NiTi wire has a high TTR, and wire deflection is taking place at low temperature, the wire will likely remain in the martensitic phase at all times during deflection and not exhibit the effects of phase transformation. In general, austenitic NiTi wires are relatively stiffer than martensitic NiTi wires (Santoro, Nicolay, & Cangialosi, 2001). To summarize, the superelasticity phenomenon of NiTi alloys is due to the transformation between the martensitic and austenitic phases, which can be induced by stress (Santoro & Beshers , 2000).

"Shape memory" is another clinically useful property of NiTi alloys. If a NiTi wire is subjected to repeated deflection and temperature cycles, the wire while in the austenitic phase will retain a preformed shape. If the temperature of the austenitic wire is lowered, the martensitic phase transformation will begin to occur, and the wire will become more pliable and able to accept deformations. As the temperature is increased, the wire will transform back towards austenite and regain its preformed shape (Santoro, Nicolay, & Cangialosi, 2001).

Nickel-Titanium Coil Springs

Miura et al. (1988) conducted one of the first studies to examine the application of the nickel-titanium alloy in the orthodontic intraoral appliance. It was known that nickel-titanium alloy had superelastic characteristics in archwires, but the force characteristics in coil springs was not well understood. In general, a superelastic nickel-titanium wire has a near constant deactivation force, and the wire can withstand relatively large deflection before permanent deformation occurs (Miura, Mogi, Ohura, & Karive, 1988). Graphically, superelastic behavior is evident as a plateau in a load-deflection graph during the deactivation phase. Miura et al. compared nickel-titanium to stainless steel open-coil springs and also examined the effects of coil wire diameter, coil lumen dimension, martensite transformation temperature variation, and coil pitch. Testing was conducted at 37^{o}C +/- 1.0 ^{o}C to simulate intraoral temperatures and compression rate was 10 mm per minute. The testing results demonstrated that the nickel-titanium open-coil springs produced near constant deactivation forces without signs of permanent deformation. The superelastic plateau occurred while the spring length was between 15-75% of pre-compression length. The stainless steel springs produced deactivation forces that increased with greater amounts of compression and became permanently deformed after maximal compression. For the nickel-titanium open-coil springs, deactivation forces increased with greater coil wire diameter, decreased with greater lumen dimension, and decreased with greater martensite transformation temperature. As pitch increased, the range where the deactivation force is near constant also increased, but overall force-level was nearly unaffected. Miura et al. keenly points out that as the percent compression of an open-coil nickel titanium spring approaches zero, the deactivation force is below the threshold for tooth movement. Clinically, this suggests the millimeters of coil incorporated in the intraoral appliance should be greater than the amount of desired tooth movement. The superelastic behavior and differences between stainless steel and nickel-titanium can primarily be explained by the stress-induced martensite transformation of nickel-titanium (Miura, Mogi, Ohura, & Karive, 1988).

A study by von Fraunhofer et al. (1993) arrived at a similar conclusion to Miura et al. (1988) regarding the greater utility of nickel-titanium in coil springs. The stainless steel springs tested produced heavy, rapidly decaying forces upon deactivation whereas the nickel-titanium springs produced lighter and relatively constant deactivation forces (von Fraunhofer, Bonds, & Johnson, 1993).

Bourke et al. (2010) and Brauchli et al. (2011) conducted studies that examined the force deflection characteristics of a series of commercially available nickel-titanium open-coil springs. The former compressed the springs to 50% of the original length and the latter compressed the springs to 25% and 50% of the original length. Bourke et al. (2010) noted the actual average deactivation forces of the springs were between 9 and 42% below the reported force labeled by the manufacturers, which included GAC International (Bohemia, NY) and 3M Unitek (Monrovia, CA). Interestingly, maximum force values were significantly higher than the labeled force value. It was noted that the manufacturer method of determining force values is not well understood (Bourke, Daskalogiannakis, Tompson, & Watson, 2010). Another interesting conclusion from this study is that as the same springs were tested multiple times over a period of 12 weeks, the average deactivation forces for the springs decreased over time. The springs were also found to display non-superelastic behavior, indicated by the high load-deflection ratios, and deactivation forces decreased as the springs decompressed.

Brauchli et al. (2011) conducted a related study, but it differed primarily by the addition of a 25% compression level. Most of the nickel-titanium open-coil springs were found to have a more linear load deflection graph, but the GAC springs (Dentsply GAC, Bohemia, NY) and the RMO springs (Rocky Mountain Orthodontics, Denver, Colorado) were found to display superelastic behavior, evidenced by a clear force plateau (Brauchli, Senn, Ball, & Wichelhaus, 2011). This superelastic behavior only occurred in the 50% compression test. It is believed the superelastic behavior did not occur in the 25%

compression test because there was not enough deformation present to create lattice transformation. Spring manufacturers have tried to mitigate this issue by reducing the number of coils in a given length of spring, so that a small amount of activation creates a relatively larger degree of deformation and subsequently inducing lattice transformation. One method for doing so was the stop-wound spring. Still, these springs did not exhibit superelastic behavior in either the Brauchli et al. (2011) or the Bourke et al. (2010) studies. The absence of superelastic behavior was speculated as attributable to the use of non-superelastic nickel-titanium alloys such as stabilized or work-hardened martensite (Brauchli, Senn, Ball, & Wichelhaus, 2011). Lastly, Brauchli et al. (2011) recommends utilizing a coil-spring that is 200% the interbracket distance clinically, which should result in spring compression upon activation of at least 50%. This should allow for the utilization of the superelastic behavior found in the GAC and RMO springs.

CHAPTER 3
MATERIALS AND METHODS

In this study, four open-coil spring products were evaluated: DENTSPLY GAC Sentalloy Coil Spring (15 mm; 100 and 150 gram coils; 0.009” x 0.030” wire size and inner diameter; Dentsply GAC, Islandia, NY), Orthoclassic Orthodontics (15 mm; 150 gram; .035” inner diameter; McMinnville, OR), and American Orthodontics (20 mm; 0.010” x 0.030” wire size and inner diameter; Sheboygan, WI) nickel-titanium springs. Sixty of each type were collected and randomly distributed into test groups of n=15. Four different compression (activation) levels were utilized with reference to the 15 mm pre-cut length of the coils so that the coils were compressed to 20, 40, 60, or 80% of original length. Testing was conducted so that each type of coil spring is tested at each of the four compression levels. Each spring is tested one time. The American Orthodontics coil springs arrived pre-cut to 20 mm. All other coils were precut to 15 mm. The length of the American Orthodontics springs was reduced so that it measured 15 mm by trimming 5 mm of spring with a wire cutter. The terminal ends of the American Orthodontics coil springs are not flat and are unfinished. All of the other coil springs have flat, finished ends.

A universal testing machine (Instron Corp, Canton, MA) applied the compression force at 10 mm/minute to achieve each distance of activation, which included 3, 6, 9, and 12 mm of compression to correspond to 80, 60, 40, and 20% of original spring length, respectively. Measurements were conducted at 37 +/- 1^{o}C. The temperature was monitored with a Temp Alert Dual Thermo air thermometer (Fisher Scientific, Hampton NH).

The experimental setup utilized a cylinder with a hole drilled through the top surface such that it allowed a 0.020 inch stainless steel guiding wire to pass through, but not the coil. The guiding wire was attached to a Jacobs chuck, which held the open-coil springs in place for testing.

Springs (n=15) of each type were compressed at each activation level. Each spring was compressed and tested only once. After testing, each spring was placed in an envelope and labeled so that each spring could be matched to its corresponding test.

Activation and deactivation forces at mm increments were compared within each product among the percent compressed groups (20, 40, 60, 80%). IBM SPSS Statistics software (Armonk, NY) was used to analyze the data with ANOVA using a significance value of $p<0.05$. Activation and deactivation forces were analyzed separately. Force versus compression curves were compared for each product.

CHAPTER 4
RESULTS

In Tables 1-8 below, the average force values and standard deviation for each coil group across the activation range are shown.

Coil Compression	1 mm	2 mm	3 mm
GAC 150 (g)	86 ± 4	181 ± 9	263 ± 12
GAC 100 (g)	84 ± 7	170 ± 10	216 ± 11
AO	55 ± 7	115 ± 11	173 ± 15
OO Medium	21 ± 1	44 ± 1	66 ± 2

Table 1. Activation forces (grams) for 3 mm total activation group

Coil Compression	2 mm	1 mm
GAC 150 (g)	164 ± 6	77 ± 3
GAC 100 (g)	139 ± 6	72 ± 6
AO	80 ± 5	38 ± 3
OO Medium	40 ± 2	20 ± 2

Table 2. Deactivation forces (grams) for 3 mm total activation group

Coil Compression	1 mm	2 mm	3 mm	4 mm	5 mm	6 mm
GAC 150 (g)	90 ± 7	187 ± 12	265 ± 20	303 ± 20	305 ± 20	330 ± 17
GAC 100 (g)	85 ± 6	179 ± 11	244 ± 13	252 ± 19	266 ± 17	288 ± 21
AO	59 ± 11	121 ± 21	181 ± 26	227 ± 22	263 ± 14	293 ± 12
OO Medium	25 ± 17	53 ± 35	77 ± 49	98 ± 53	116 ± 48	133 ± 51

Table 3. Activation forces (grams) for 6 mm total activation group

Coil Compression	5 mm	4 mm	3 mm	2 mm	1 mm
GAC 150 (g)	201 ± 13	200 ± 12	206 ± 11	159 ± 7	77 ± 5
GAC 100 (g)	157 ± 19	150 ± 20	155 ± 17	150 ± 9	74 ± 4
AO	193 ± 11	158 ± 16	123 ± 23	84 ± 22	41 ± 11
OO Medium	96 ± 30	82 ± 35	67 ± 37	47 ± 32	23 ± 16

Table 4. Deactivation forces (grams) for 6 mm total activation group

Coil Compression	1 mm	2 mm	3 mm	4 mm	5 mm	6 mm	7 mm	8 mm	9 mm
GAC 150 (g)	87 ± 6	182 ± 11	263 ± 13	302 ± 18	301 ± 31	336 ± 33	370 ± 34	399 ± 20	419 ± 18
GAC 100 (g)	83 ± 5	175 ± 9	234 ± 12	247 ± 18	253 ± 14	272 ± 22	291 ± 21	217 ± 23	330 ± 26
AO	56 ± 12	118 ± 22	179 ± 28	225 ± 23	256 ± 16	291 ± 18	320 ± 18	346 ± 22	373 ± 22
OO Medium	25 ± 17	52 ± 34	77 ± 48	98 ± 53	114 ± 41	133 ± 48	149 ± 54	163 ± 62	189 ± 60

Table 5. Activation forces (grams) for 9 mm total activation group

Coil Compression	8 mm	7 mm	6 mm	5 mm	4 mm	3 mm	2 mm	1 mm
GAC 150 (g)	203 ± 17	186 ± 19	183 ± 12	181 ± 10	185 ± 8	200 ± 11	150 ± 7	71 ± 4
GAC 100 (g)	167 ± 11	155 ± 11	146 ± 12	143 ± 14	143 ± 15	148 ± 14	143 ± 6	69 ± 6
AO	223 ± 26	210 ± 22	193 ± 17	173 ± 13	144 ± 16	113 ± 22	76 ± 22	35 ± 11
OC Medium	107 ± 20	97 ± 19	92 ± 24	85 ± 26	74 ± 31	60 ± 36	42 ± 31	19 ± 14

Table 6. Deactivation forces (grams) for 9 mm total activation group

Coil Compression	1 mm	2 mm	3 mm	4 mm	5 mm	6 mm
GAC 150 (g)	87 ± 6	181± 10	263 ± 16	303 ± 19	315 ± 22	335 ± 19
GAC 100 (g)	83 ± 4	168 ± 6	211 ± 6	215 ± 12	226 ± 16	245 ± 12
AO	54 ± 7	112 ± 10	174 ± 22	221 ± 12	258 ± 12	292 ± 12
OO Medium	21 ± 1	43 ± 2	65 ± 2	85 ± 2	103 ± 3	121 ± 3
Coil Compression	7 mm	8 mm	9 mm	10 mm	11 mm	12 mm
GAC 150 (g)	361 ± 25	392 ± 29	408 ± 25	419 ± 17	435 ± 16	450 ± 16
GAC 100 (g)	261 ± 8	280 ± 11	297 ± 16	317 ± 13	324 ± 13	324 ± 18
AO	320 ± 17	346 ± 18	373 ± 20	398 ± 22	424 ± 28	678 ± 43
OO Medium	136 ± 4	150 ± 6	171 ± 10	203 ± 15	241 ± 16	272 ± 19

Table 7. Activation forces (grams) for 12 mm total activation group

Coil Compression	11 mm	10 mm	9 mm	8 mm	7 mm	6 mm
GAC 150 (g)	257 ± 22	233 ± 17	200 ± 15	186 ± 9	178 ± 8	177 ± 7
GAC 100 (g)	169 ± 9	151 ± 10	133 ± 9	124 ± 6	119 ± 6	116 ± 5
AO	271 ± 24	244 ± 20	223 ± 18	217 ± 17	201 ± 16	187 ± 11
OO Medium	130 ± 9	110 ± 8	98 ± 9	91 ± 8	86 ± 6	80 ± 4
Coil Compression	5 mm	4 mm	3 mm	2 mm	1 mm	
GAC 150 (g)	177 ± 6	177 ± 7	191 ± 10	143 ± 5	64 ± 6	
GAC 100 (g)	113 ± 5	113 ± 6	114 ± 8	120 ± 7	66 ± 7	
AO	167 ± 9	136 ± 11	104 ± 8	68 ± 6	31 ± 5	
OO Medium	71 ± 3	59 ± 3	44 ± 3	28 ± 2	10 ± 2	

Table 8. Deactivation forces (grams) 12 mm total activation group

Table 9 demonstrates the mean force value at 3 mm compression upon deactivation for each coil brand and for all four levels of initial activation.

Activation	3mm (80%)	6mm (60%)	9mm (40%)	12mm (20%)
GAC 150 (g) Mean	174	123	113	104
GAC 100 (g) Mean	216	155	148	114
AO Mean	263	206	200	191
OO Medium Mean	66	66	60	44
Total Mean	180	137	130	113

Table 9. Mean deactivation force (grams) at 3 mm compression upon deactivation

The following figures (Figs. 1-21) demonstrate graphically the forces generated during each of the activation/deactivation ranges for each group. The black and white figures represent an example of one coil from each activation group for each brand as listed. The colored figures represent a compilation of the four graphs from a given test group for each brand as listed.

For all four open-coil spring products, deactivation force values at 3 mm significantly ($p<0.05$) decreased with greater activation/compression level. When activated to 20% of original length, the force values at 3 mm compression upon deactivation were 53-73% of the force level when activated to 80% of original length. When activated to 40% of original length, the force values at 3 mm compression upon deactivation were 65-91% of the force level when activated to 80% of original length.

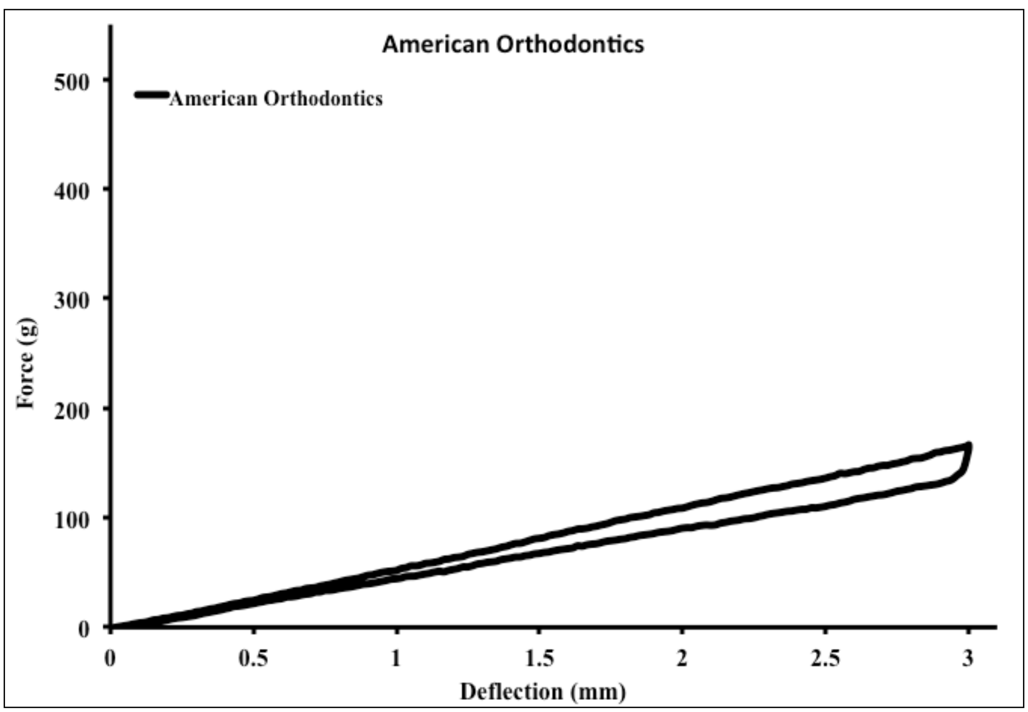

Figure 1. Graph of AO coil 3 mm compression

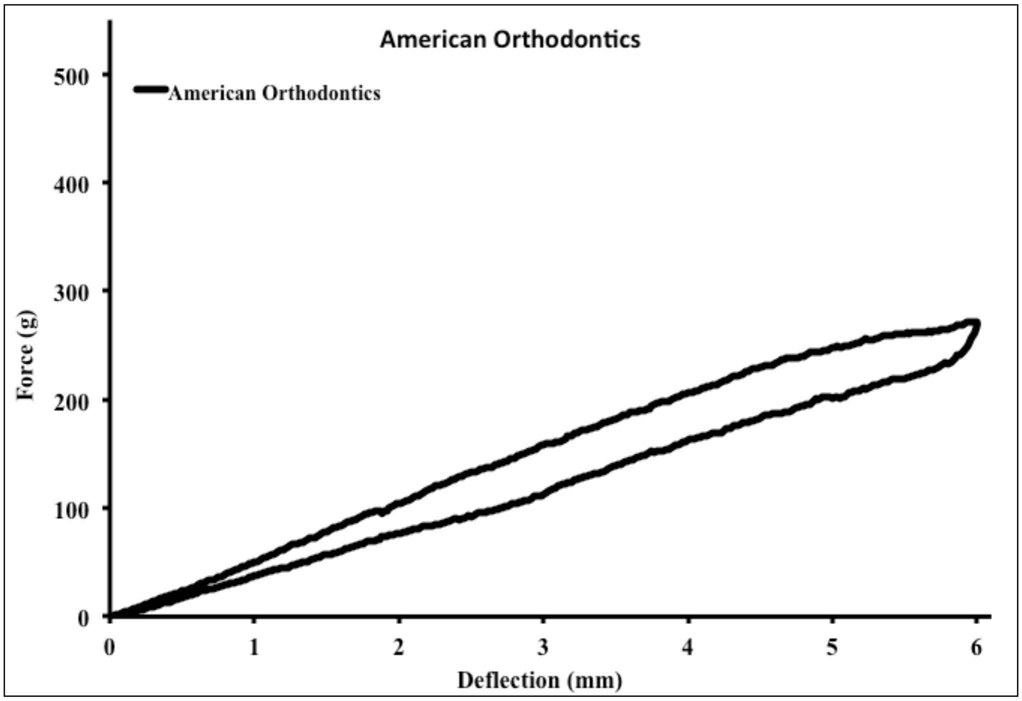

Figure 2. Graph of AO coil 6 mm compression

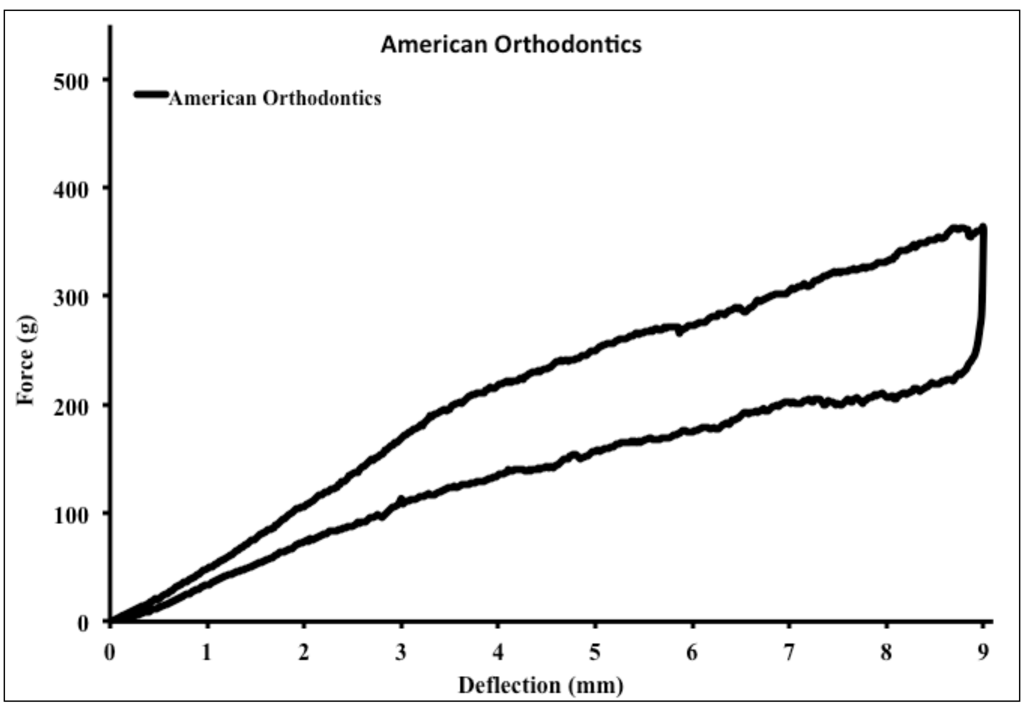

Figure 3. Graph of AO coil 9 mm compression

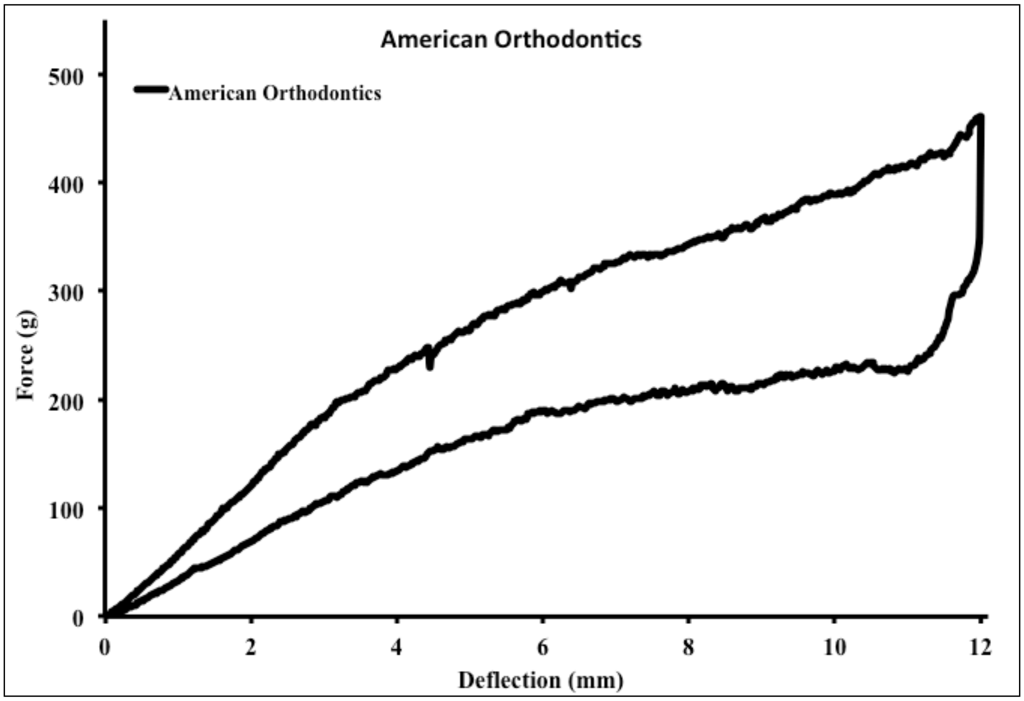

Figure 4. Graph of AO coil 12 mm compression

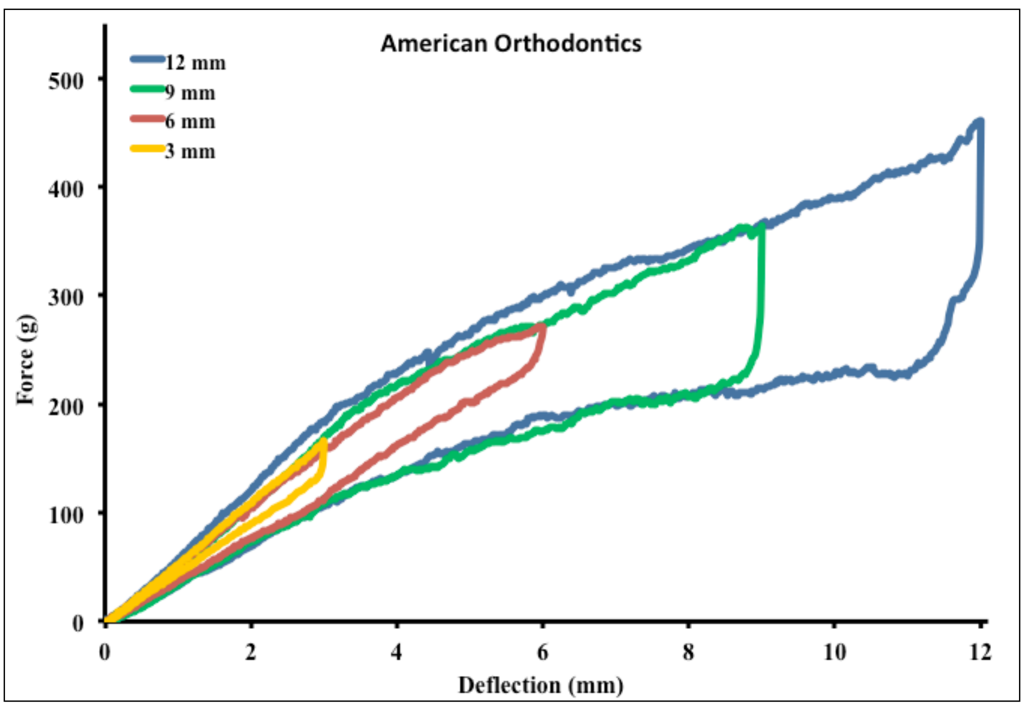

Figure 5. Composite graph of all four AO coil compression levels

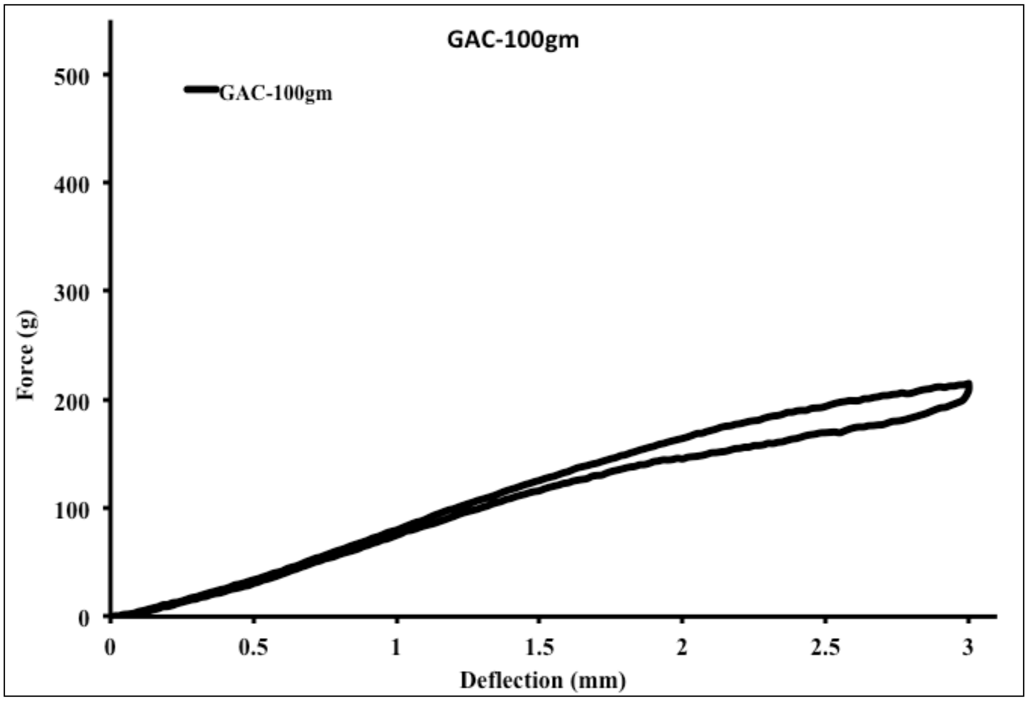

Figure 6. GAC-100gm coil 3 mm compression

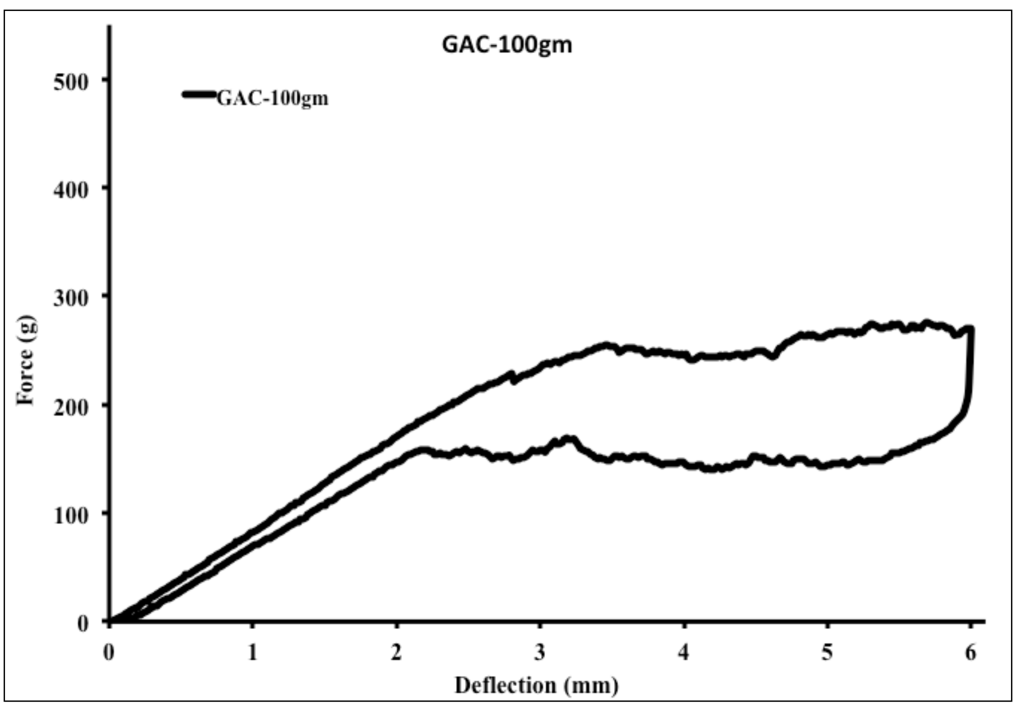

Figure 7. GAC-100gm coil 6 mm compression

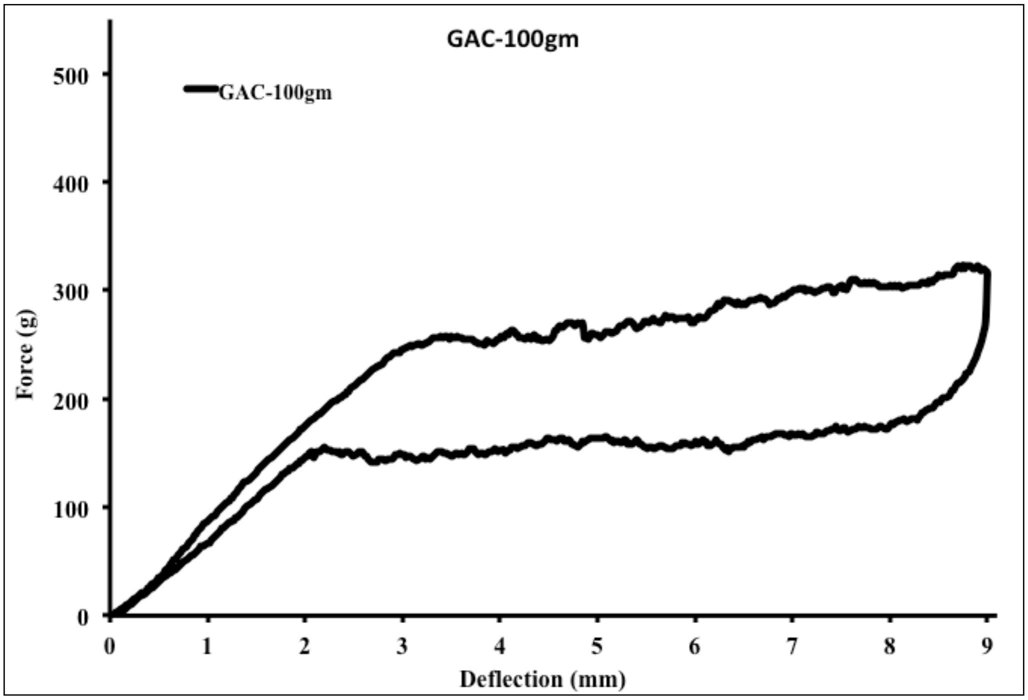

Figure 8. GAC-100gm coil 9 mm compression

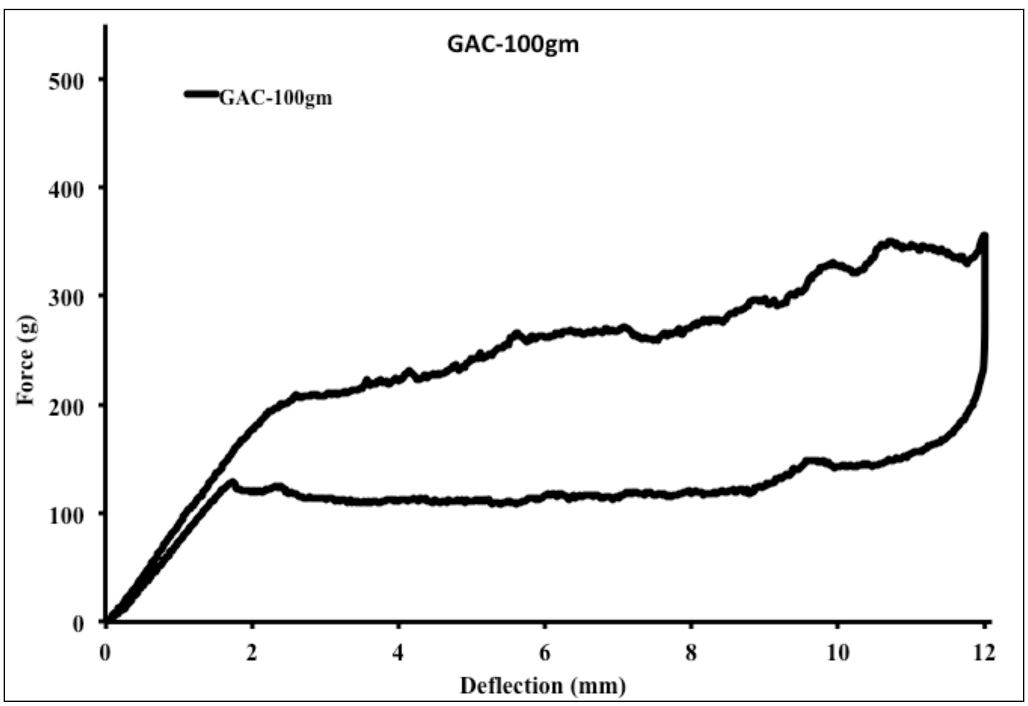

Figure 9. GAC-100gm coil 12 mm compression

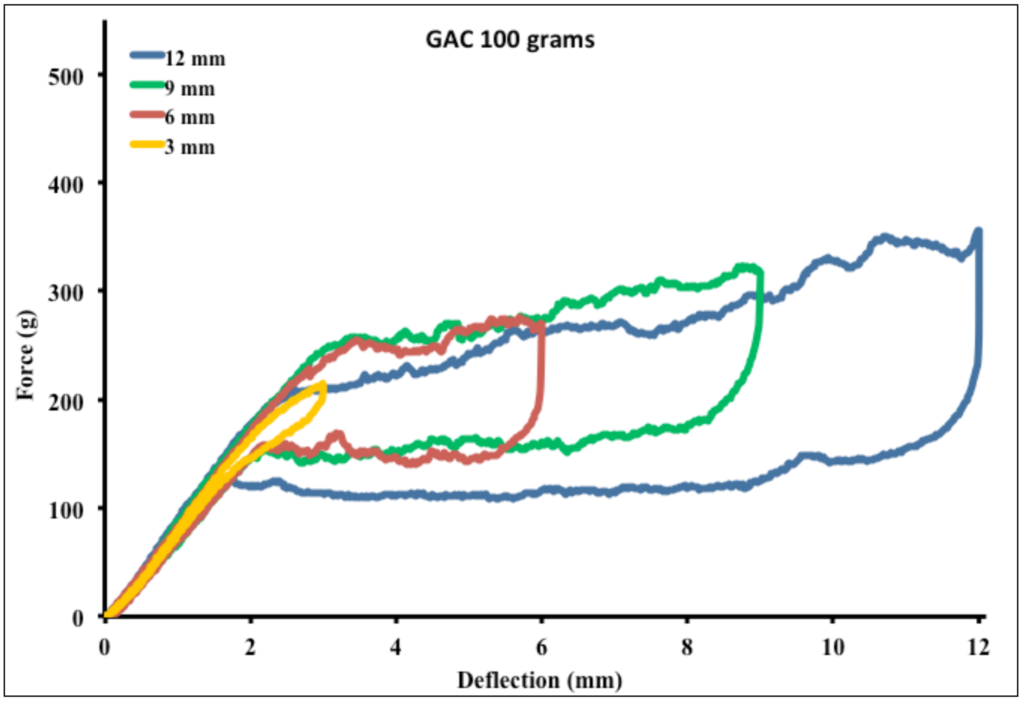

Figure 10. GAC-100gm composite of all four compression levels

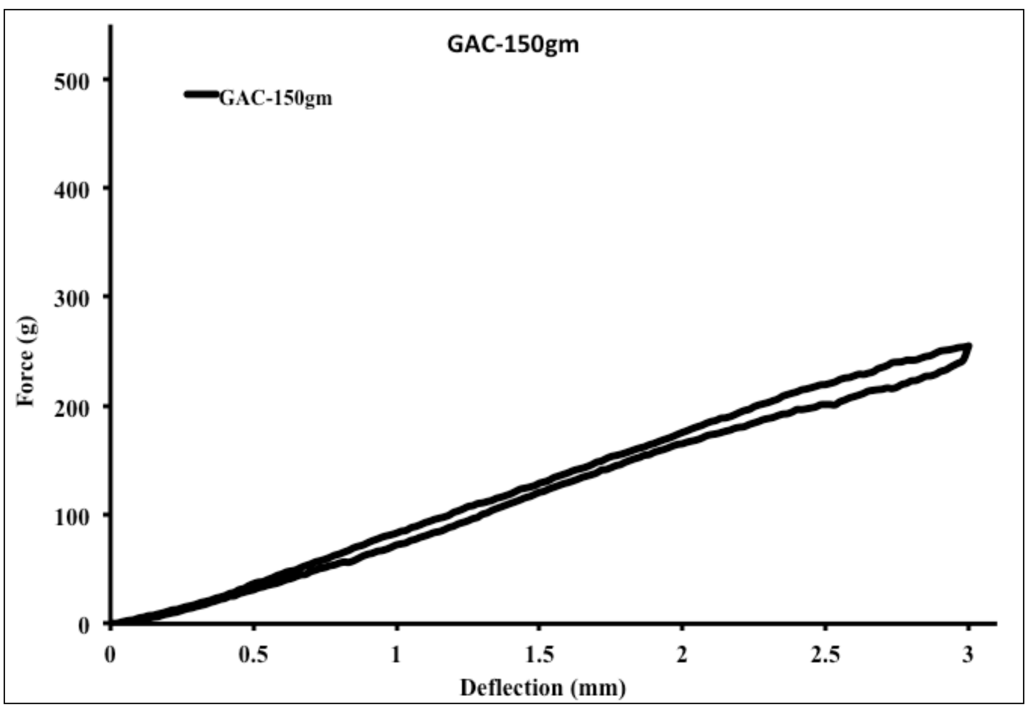

Figure 11. GAC-150gm 3 mm compression

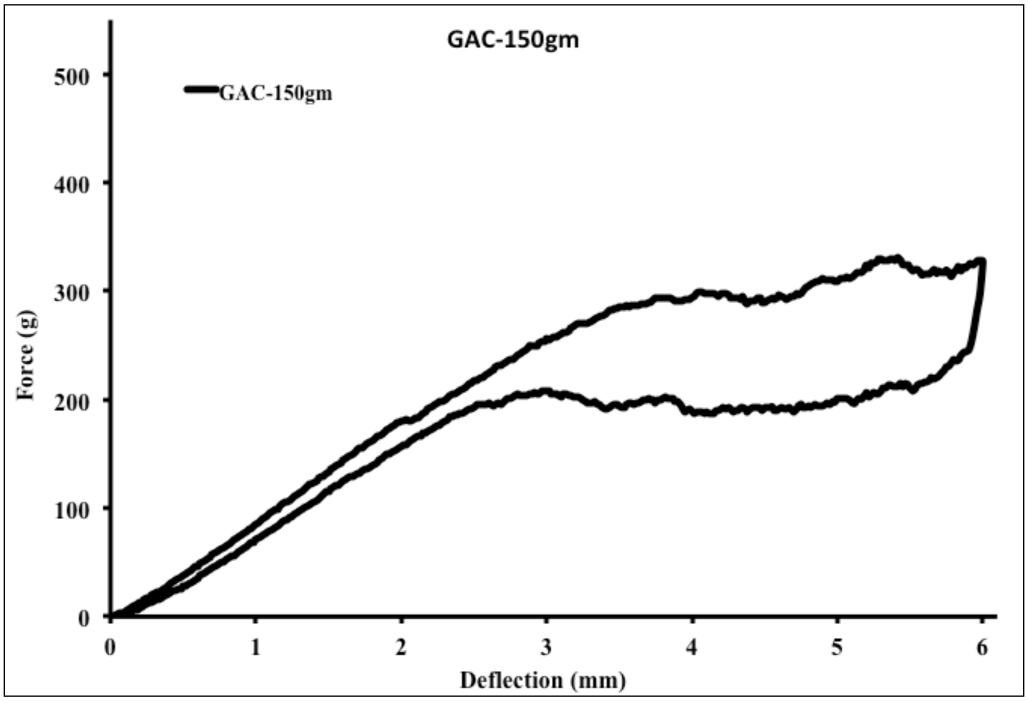

Figure 12. GAC-150gm 6 mm compression

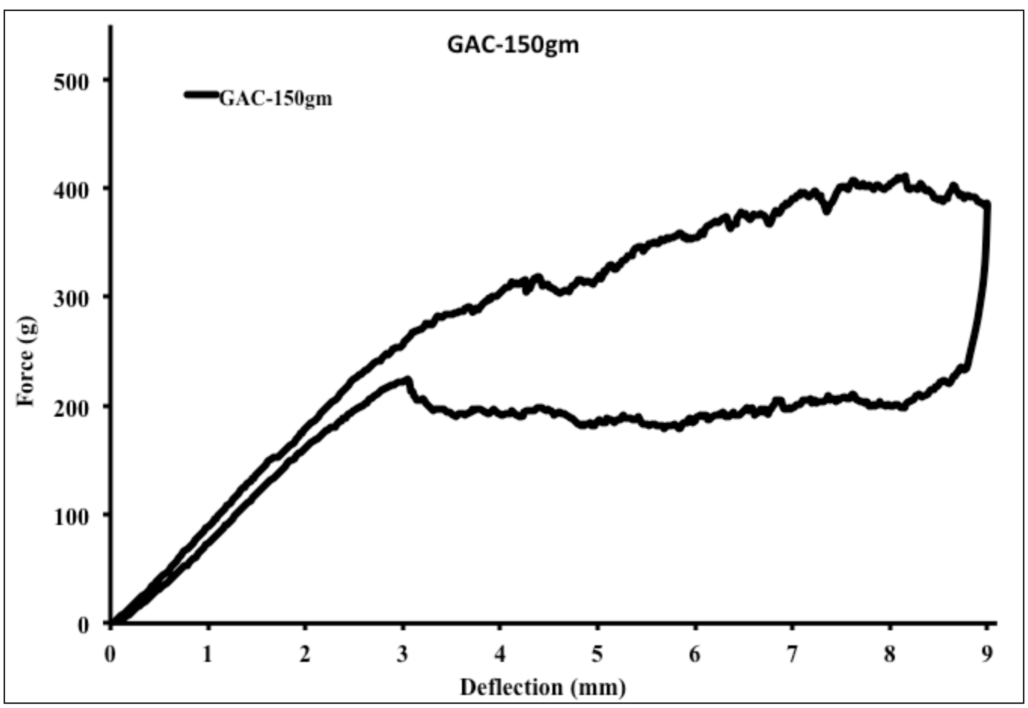

Figure 13. GAC-150gm 9 mm compression

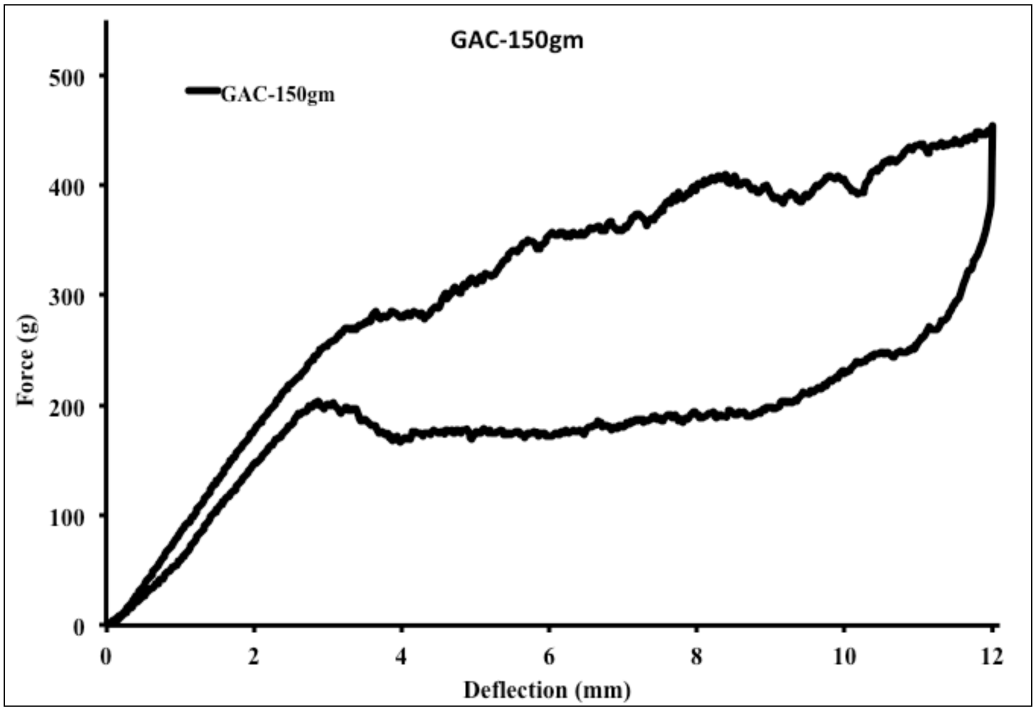

Figure 14. GAC-150gm 12 mm compression

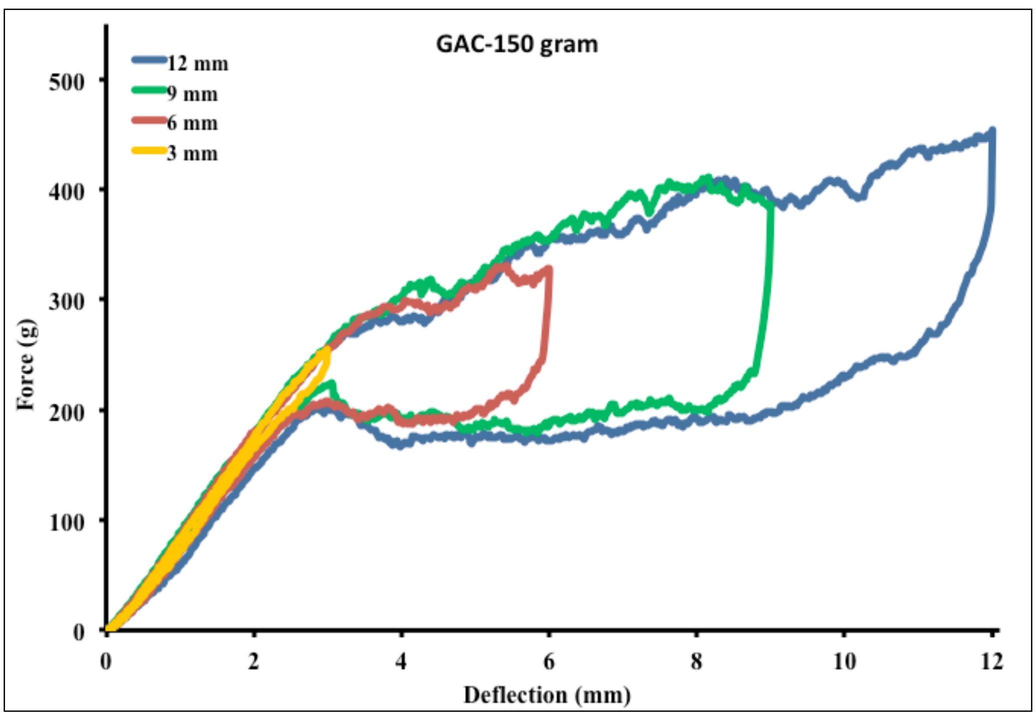

Figure 15. GAC-150gm composite all four compression levels

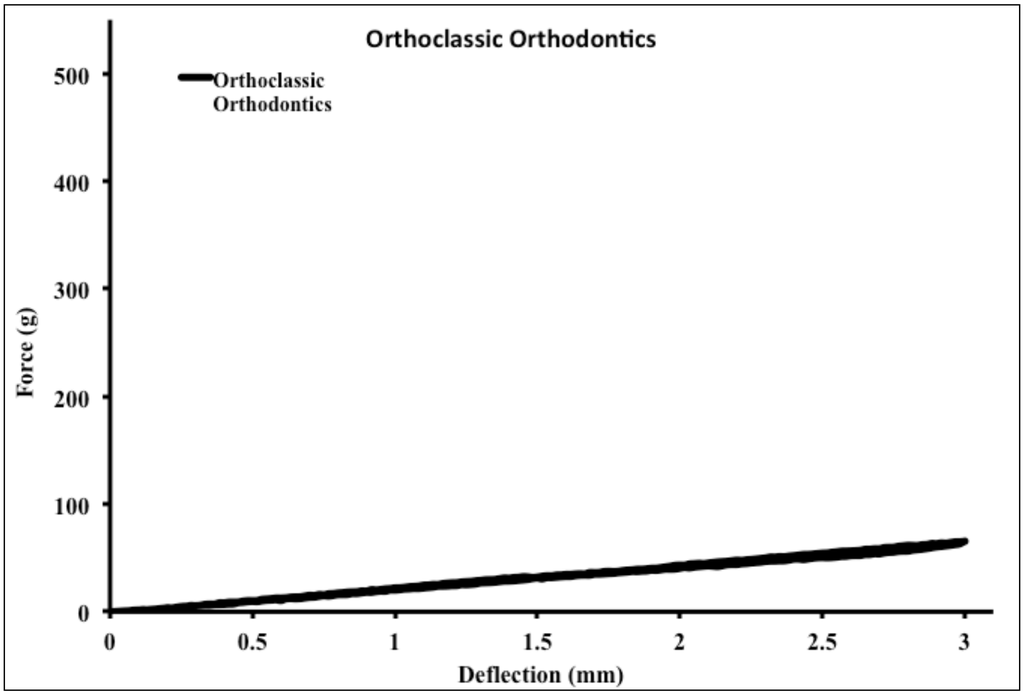

Figure 16. OO coil 3 mm compression

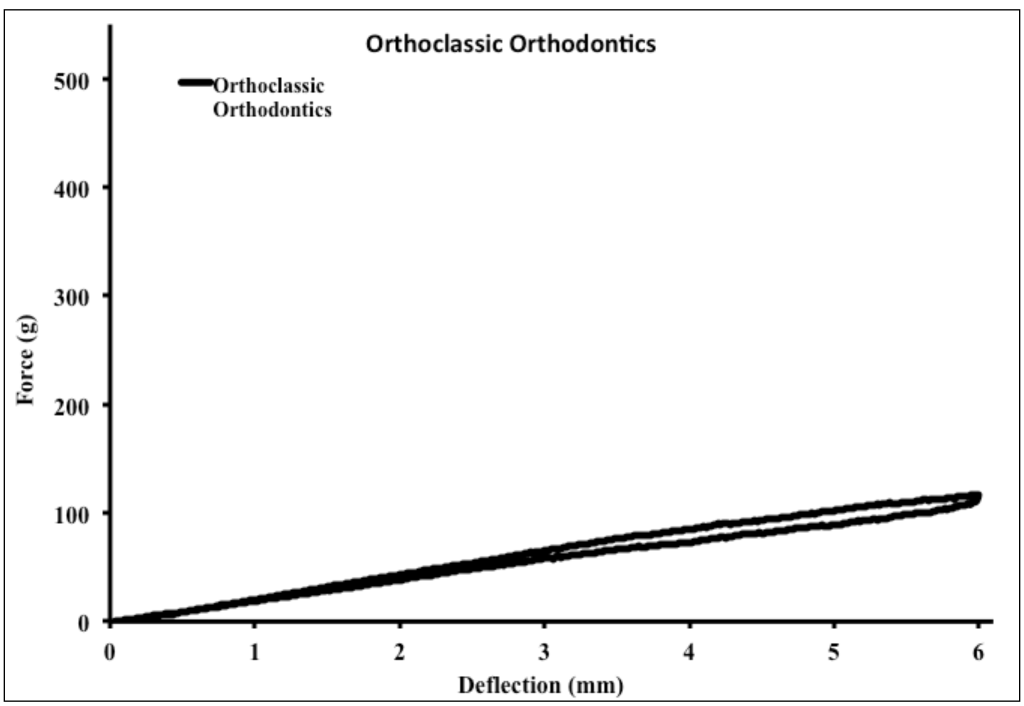

Figure 17. OO coil 6 mm compression

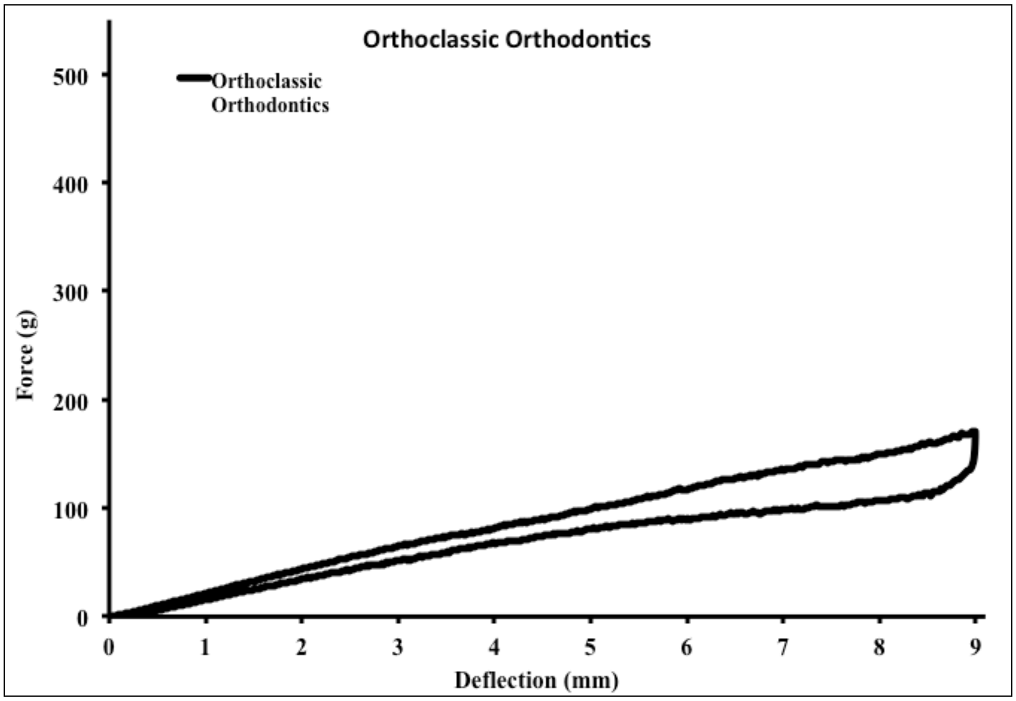

Figure 18. OO coil 9 mm compression

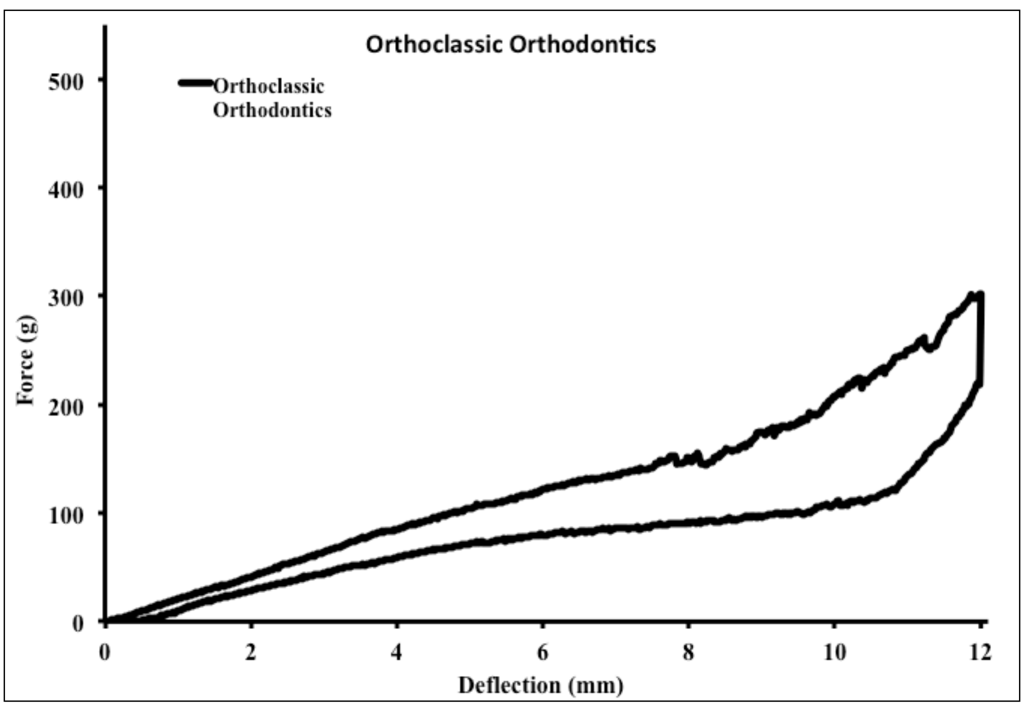

Figure 19. OO coil 12 mm compression

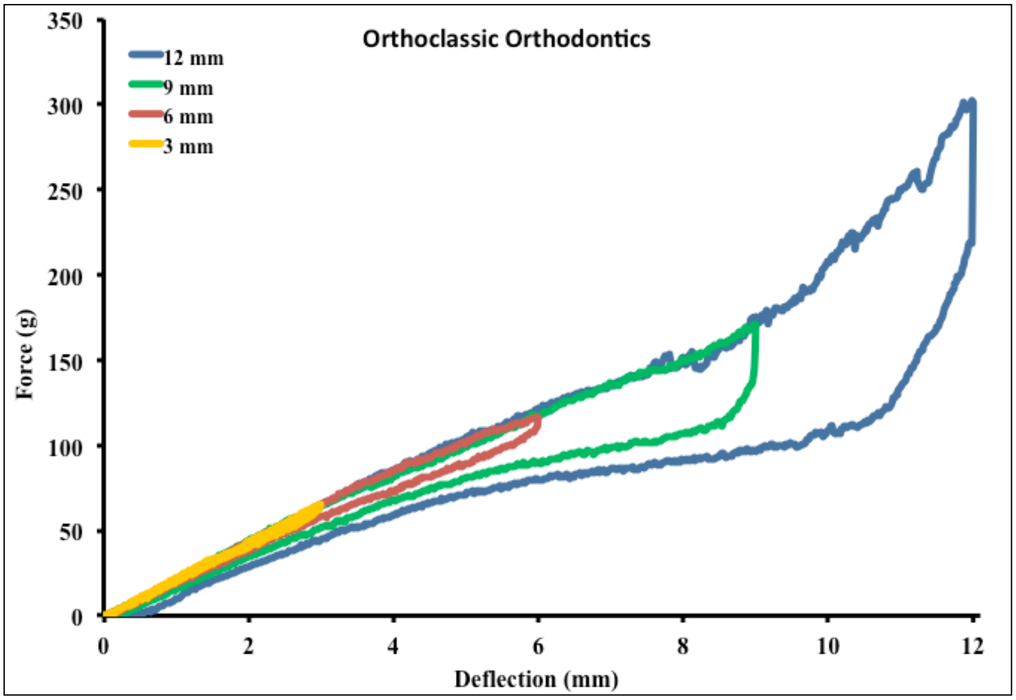

Figure 20. Composite graph of all four Orthoclassic

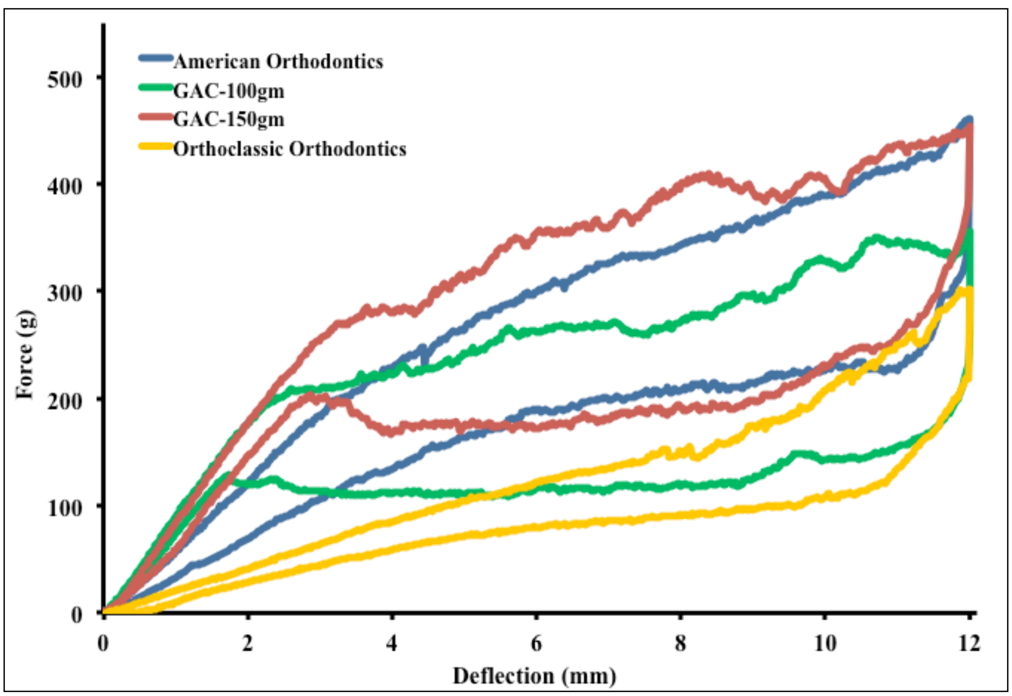

Figure 21. Comparison of AO, GAC, and OO coils at 12 mm compression

CHAPTER 5
DISCUSSION

Nickel-titanium open-coil springs are a routinely used and effective auxiliary for the orthodontic intraoral appliance. Studies have shown repeatedly that nickel-titanium open-coil springs are capable of producing light, continuous forces when properly utilized. In this study, the deactivation forces of four commercially available nickel-titanium open-coil springs were analyzed using four different levels of initial activation.

Prior to beginning the study, it was hypothesized that as the amount of initial activation of a nickel-titanium open-coil spring increases, the resultant deactivation force at a given level of compression upon deactivation will also increase. The data collected do not support this hypothesis. As seen in Table 9, as the amount of initial activation increased from 3 mm to 12 mm (or activation from 80% of original length to 20% of original length), the average deactivation force at 3 mm compression upon deactivation significantly decreased ($p<0.05$) for each coil type. 3 mm compression upon deactivation was utilized to compare the deactivation forces because a deactivation force could be recorded at this level for each test group. A likely explanation for this trend is related to the absence of a superelastic plateau observed in the 3 mm initial activation test group of the GAC 100 gram and 150 gram springs. As the amount of initial coil activation increases, there is increased deformation of the coil wire which will likely lead to the production of stress-induced martensite. This phase transformation from austenite to martensite resulted in the lower deactivation force levels.

As visualized in Figure 1-21, the degree of superelastic behavior differed between test groups. A clear superelastic force plateau occurred during deactivation for both the

100 gram and 150 gram GAC coil groups with initial activations of 6 mm, 9 mm, and 12 mm. The GAC test groups with 3 mm initial activation did not exhibit a superelastic force plateau. The AO and OO test groups, between all ranges of initial activation, exhibited a more linear load deflection graph. A possible explanation for the lack of superelastic force plateau in the AO and OO springs is a difference in NiTi material properties between the GAC springs and AO/OO springs. Additionally, the reported GAC coil-wire diameter was 0.001 inches smaller than the AO and OO coil-wire diameters. It is likely that a superelastic force plateau was not seen in the GAC groups with 3 mm initial activation because the spring was only activated to 80% of initial length, which is likely insufficient for the austenite to martensite phase transformation to occur. This is supported by the findings of Brauchli et al. (2011) who also tested GAC springs and found superelastic force plateaus in activation to 50% of original length but not in activation to 75% of original length. It was suggested that there was insufficient deformation produced in the latter group to cause lattice transformation (Brauchli, Senn, Ball, & Wichelhaus, 2011).

For the GAC springs that exhibited a clear superelastic force plateau, a greater initial activation lowered the force level at which the plateau occurred. For example, as seen in Figure 7 and 9 for the GAC 100 gram test groups, the superelastic plateau occurred at approximately 110 grams of force for the 12 mm compression test group and approximately 150 grams of force for the 6 mm compression test group. This phenomenon is referred to as hysteresis.

Two of the three coil-spring brands tested, GAC and OrthoClassic, provided information on the force levels of their products. The force levels of the GAC coils were reported as 100 grams and 150 grams, and the force levels of the OrthoClassic coils were reported as 150 grams. This information can be valuable to clinicians as many may not have the equipment needed to identify force levels of coil springs, and different clinical applications require different force levels. These brands, however, do not provide details on recommended activation levels or the degree of activation required to produce the reported force levels. The superelastic force plateaus for the GAC springs are generally higher than the listed force levels for GAC springs, as seen in Figure 10 and 15. The tested force levels during superelastic plateaus begin to approach the listed force levels with greater initial activation. For OrthoClassic, the listed force levels are generally higher than tested force levels, which only reached the listed levels during maximal activation and not during deactivation. As seen in Table 8, the highest recorded mean of deactivation forces for the OrthoClassic Medium (150 gram) spring was 130 grams during the 12 mm initial activation test group.

Brauchli et al. (2011) recommended clinicians activate nickel-titanium open-coil springs to at least 50% of initial length to produce continuous forces after finding a superelastic force plateau only with compression to 50% of initial length and not with compression to 75% of initial length. This can be accomplished by utilizing a spring that is 200% of the interbracket distance where the spring will be placed (Brauchli, Senn, Ball, & Wichelhaus, 2011). Without sufficient activation and deformation to induce phase transformation, the resultant forces on the teeth will likely be more variable. Based on the findings and the specific size and brands of coils tested in this study, coil activation to 60% of the initial length will also be sufficient to produce continuous forces, but only with the GAC 100 and 150 gram springs. Other springs not tested in this study may produce similar patterns of deactivation forces. The clinician should also be aware that, in general, greater initial activation generates lower deactivation forces. This may seem counterintuitive to a clinician not familiar with nickel-titanium load-deflection patterns.

Placing nickel-titanium open-coil springs that are 200% of an interbracket distance is not without potential negative side effects. A coil spring will continue to apply forces to the teeth until the coil is removed or the coil spring has deactivated to its initial length. Consequently, if a fixed intraoral appliance fitted with an open-coil spring sized to 200% the interbracket distance is left in place for an extended period of time unsupervised, the spring could displace teeth outside the bony housing, especially if utilizing a continuous, low-stiffness, NiTi archwire. If a spring is being used for distalization of a posterior tooth, an unmonitored open-coil spring could cause excess tooth movement or impaction of other unerupted teeth. A myriad of other negative consequences are possible. Therefore, it is essential for the clinician to closely monitor patients with coil-springs compressed more than the desired total amount of tooth movement. This remains true for many of the intraoral appliances available in the orthodontists' armamentarium.

CHAPTER 6
CONCLUSION

Nickel-titanium open-coil springs exhibit lower deactivation forces with greater amounts of initial activation. Superelastic behavior was observed in the form of a superelastic plateau in the GAC 100 gram and 150 gram springs with initial activation levels of 6 mm, 9 mm and 12 mm. A lesser degree of superelastic behavior was observed in the AO and OO springs. Based on these results, nickel-titanium open-coil springs appear to be an effective orthodontic auxiliary at producing continuous forces, especially when activated by compression to at least 60% of original length in certain brands of coil springs.

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