Influence of Taper on the Flexibility of Nickel-Titanium Rotary Files

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INFLUENCE OF TAPER ON THE FLEXIBILITY OF NICKEL-TITANIUM ROTARY FILES

by

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ABSTRACT

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Introduction: Modern nickel-titanium instruments have various tapers and have been marketed to have superior flexibility from previous generations. Current ISO standards require force measurements at a static point along the file. Unfortunately, root canal anatomy varies and produces multiple forces along the length of the file. The purpose of this study was to determine the influence of taper on the flexibility of various nickel-titanium files.

Materials and Methods: The flexibility of stainless steel hand files and nickel-titanium rotary files of various tapers was measured. The sample size was 10 for each type, taper and size. The files were measured at 3, 5 and 7 mm from the tip using a digital caliper and marked with a rubber stopper and a distance of 20mm from the tip was used as the deflection point. Each file was securely fastened on a load-sensing cell and bending was accomplished using a universal testing machine to a maximum deflection of 4.5 mm at a rate of 2 mm/minute under room temperature conditions ($22^{O}C \pm 1^{O}C$). Data was collected electronically via Merlin Software and transferred to Microsoft Excel. Statistical analysis was completed with IBM SPSS Statistics software and a two-way analysis of variance (ANOVA) was used as well as a Post-hoc Tukey test.

Results: The force and bending moments of EndoSequence .06 taper files was significantly greater ($p<0.05$) than EndoSequence .04 and stainless steel hand files at all lengths. No significant difference was noted between EndoSequence .04 and stainless steel hand files from 0.25 mm to 3.0 mm. From 3.5 mm to 4.5 mm, the force and bending moments for stainless steel hand files was significantly greater ($p< 0.05$) than EndoSequence .04 files. Within each file group, the force and bending moments were significantly greater ($p<0.05$) as the grasp length increased (7 mm>5 mm>3 mm).

Conclusions: With a vast array of root canal instruments currently available clinicians should consider the properties of instruments before cleaning and shaping. Nickel-titanium files with tapers greater than 0.04 should not be used for apical enlargement of curved canals because these files are significantly stiffer resulting in an increased chance of canal transportation.

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INTRODUCTION

Non-surgical root canal therapy may be defined as the chemo-mechanical preparation of the root canal system followed by three-dimensional filling with an inert material to restore or maintain the health of the periradicular tissue [1]. Kakehashi et al.'s landmark study in 1965 was the first to definitively show that the presence of bacteria in the root canal system leads to pulpal pathology and periapical breakdown [2]. Based on these findings, the fundamental objective of endodontic treatment is to prevent or resolve apical periodontitis through disinfection and sealing of the root canal system [3].

While sealing of the root canal system is important, it has long been proven that the most important phase of endodontic treatment is cleaning and shaping of the root canal system [1]. Schilder stated that root canal instrumentation should debride the canal of pulpal tissue, remove microbes and affected dentin as well as prepare the canal for obturation [1]. Although research has shown that mechanical instrumentation alone can effectively remove bacteria from the root canal system thorough chemo-mechanical preparation can further reduce the number of microorganisms by 100 to 1000 times that of mechanical instrumentation alone [4, 5].

Recent evidence suggests that the preparation shape (i.e. shaping) and disinfection (i.e. cleaning) are interdependent steps that are intimately related [6]. Cleaning can only be effectively completed after canals have been sufficiently enlarged in the apical segment to allow passive irrigation to facilitate disinfection [7]. Conversely, canal preparation is optimized when mechanical aims are fulfilled and enlargement is acceptable to allow proper sealing [8]. The aim of modern instrumentation techniques involves enlarging the apical third of the root canal system to allow for proper debridement, disinfection and sealing of the canal space while maintaining the original root canal anatomy. Complex canal anatomy provides challenges for clinicians during mechanical instrumentation that may prevent adequate disinfection of the root canal system. Complications such as transportation, perforation, ledging and instrument separation often occur in the presence of complex canal anatomy [8-10]. Various instrumentation techniques have been developed to overcome these challenges including passive step-back, the step-down technique, crown-down pressure less technique and the balanced force technique [11-14]. Despite advances in instrumentation patterns, Weine et al. (1975) concluded that prepared canals showed characteristics reflecting the inability to maintain the general shape of the curved canal [10].

Lim et al speculated that a more flexible instrument would negotiate the complex apical anatomy easier thereby eliminating procedural errors such as apical transportation and zipping [15]. In 1988, Walia et al. introduced nickel-titanium instruments and suggested that they were three times more flexible than traditional stainless steel instruments [16]. While the increased flexibility of nickel-titanium instruments allowed operators to negotiate canal curvatures with greater ease, Camps et al. proved that nickel-titanium instruments actually presented lower torque values at failure than stainless steel instruments, resulting in a higher incidence of instrument separation [17]. Furthermore, it was determined that nickel-titanium instruments had less cutting efficiency than stainless steel instruments [18]. As a result, manufacturers began introducing nickel-titanium files with greater taper to increase the cutting efficiency and reduce torsional fatigue. While these changes resulted in more efficient instruments, the stiffness of the instruments increased. Consequently, clinicians observed more canal transportation following the outer aspect of the curvature in the apical region of root canals [19, 20].

In this study, the flexibility of stainless steel hand files (Roydent Dental Products, Johnson City, TN) and nickel-titanium rotary files (EndoSequence, Brassler USA® Dental, Savannah, GA) of various tapers was measured. Current ISO standards (ISO 3630-1) measure the resistance to bending (i.e. flexibility) of root canal instruments (stainless steel or nickel-titanium) by fixing the instrument 3mm from its tip and bending it [21]. The bending moment is measured when the angular deflection reaches 45° [21]. While this approach is acceptable it does not account for instruments with greater taper and the effect they have as canal curvature changes. The purpose of this study was to measure the bending moment of root canal instruments of different tapers at three different points along each file to a maximum deflection of 4.5 mm.

LITERATURE REVIEW

Root canal instruments date back to the year 1746 when Fouchard used annealed piano wire to make instruments by hand [22]. Instrument design continued to evolve through the nineteenth century resulting in the first commercially available intracanal instrument being introduced in 1875 [23]. Hess' seminal paper in 1921 was the first to demonstrate the need for sophisticated root canal instruments due to the complexity of root canal anatomy [24]. The importance of proper chemo-mechanical preparation was described by Stewart in 1955 when he divided root canal therapy into three distinct phases; chemo-mechanical preparation, microbial control and obturation of the root canal [25]. Stewart concluded that while each phase was important for eventual healing of the supporting tissues, chemo-mechanical canal preparation was found to be the most important. Stewart noted that as the root canal was enlarged the number of microorganisms present in the canal was reduced as well as the debris that harbors their growth [25].

Cleaning and shaping as Schilder first described in 1974 was not possible with these primitive instruments as they were designed to simply remove debris and facilitate placement of intracanal medicaments with little attempt to address the biological and mechanical demands of the root canal system. Similar studies at that time illustrated the variations in apical morphology and concluded that all canals must be thoroughly cleaned of pulpal tissue to achieve healing [26].

Based on these anatomical findings a vast array of manual root canal instruments were advocated. Kerr Manufacturing (Romulus, MI) was the first company to commercially produce root canal instruments when they introduced the K-file, K-reamer and the H-type instrument [8] K-type files and K-type reamers were manufactured by grinding square or triangular cross sections into stainless steel blanks and rotated to create a spiral shape on the file's working surface. H-type instruments were ground from tapered blanks producing a single continuous flute [8].

While the understanding of proper biomechanical objectives had drastically improved, a standardized system of instruments did not yet exist. Using microscopes, Green et al. found that commercially available instruments lacked uniformity resulting in inconsistent mechanical preparation of the root canal system [27, 28]. As a result, Ingle proposed a method of standardizing root canal instruments in 1958 that was later accepted by the Second International Conference on Endodontics and subsequently by the American Association of Endodontists in 1962 [29].

With the standardization came a great deal of research relating to mechanical properties of root canal instruments. Contrary to popular belief, Craig et al reported in 1961 that stainless steel instruments were more resistant to fracture when bending and twisting than carbon steel instruments [30]. In 1964, Sargent et al established the relationship between the cross-sectional design of stainless steel hand instruments and their resistance to fracture [31]. Camps and Pertot corroborated these findings nearly 30 years later when they concluded that the flexibility of stainless steel hand instruments was greatly influenced by cross-sectional design [32]. Specifically, stainless steel hand instruments with a square cross section were significantly less flexible than stainless steel instruments with a triangular cross section [32, 33]. Oliet and Sorin first defined what is more commonly referred to as stress-strain curves when they developed an apparatus attached to an Instron machine to measure the torsional properties of root canal instruments [34]. By providing a vertical force they were able to determine that the torsional deformation of instruments was determined by the amount of shaft rotation [34].

A great deal of research was conducted in the 1960s and 1970s to better understand the reason for instrument failure. Harty and Sondoozi reported that complete standardization of endodontic instruments was inadequate and suggested detailed specifications such as shaft design, materials, flexibility, colors, etc. were needed [35]. In 1974, the Federation Dentaire Internationale and the International Standards Organization developed new standards for root canal instruments. The American Dental Association followed suit in 1976 when the Council on Dental Materials and Devices established new standards (No. 28) for root canal instruments [36].

The main objectives of cleaning and shaping are to debride the canal of tissue, eradicate the canal of microorganisms as well as provide a uniform taper while maintaining the original canal shape [1]. The literature has shown that root canal systems need to be enlarged sufficiently (i.e. between size #35 and #40 file) to remove debris and to allow for proper irrigation in the apical third of the canal [37-42]. Unfortunately, canal curvatures in the apical region prove the most difficult in attempting to maintain the original root canal anatomy during root canal instrumentation while providing adequate space for passive irrigation. Complications such as canal transportation, zipping, ledging and instrument separation are common procedural errors that occur when instrumenting apical curvatures [10]. Such procedural errors are a direct result of the instruments used during cleaning and shaping [10]. Specifically, the stiffness of the stainless steel alloys used to manufacture root canal instruments provided limitations when negotiating canal curvatures and more importantly, a significant increase in the stiffness of the instruments was seen as the size of the instrument increases [43]. In order to limit procedural errors clinicians and manufacturers adopted numerous methods to overcome the unfavorable mechanical properties of stainless steel alloys when negotiating curved canals.

The first hand instrumentation technique was thought to be the *standardized technique* [13]. The standardized technique implements the same working length for all instruments and relies on the shape of each hand instrument to impart the final shape to the canal [13]. Given the minimal taper (0.02 mm) of the hand instruments currently available, Allison et al concluded that adequate shaping and obturation proved difficult [44]. Realizing the deficiencies of the standardized technique, Weine adapted the step-back technique, which involves a stepwise reduction of the working length in 0.5mm to 1.0mm increments with progressively larger instruments resulting in larger tapered canal preparations [10]. Walton validated the efficacy of the step-back technique in 1976 when he compared the effectiveness of filing, reaming and step-back technique and found that histologic sections showed that step-back filing was significantly more effective than filing and reaming [13]. In order to manage more difficult canal anatomy including canal curvatures and dilacerations with minimal procedural errors, Roane et al. introduced the balanced force technique in 1985 [45].

Another technique that was developed in the early 1980s is considered the step-down technique. The step-down technique promoted the shaping of the coronal 2/3 of the canal space followed by apical instrumentation [11]. The main advantage of this technique was the apical instruments were unimpeded through most of their length, which facilitated greater control, and less chance of transportation near the apical constriction [8]. Many adaptations of the step-down technique were developed including the popular crown-down pressure less technique by Marshall et al. in 1980 [8]. Morgan and Montgomery's study in 1984 validated Marshall's findings that the crown-down pressure less technique was an effective method of instrumenting curved canals. They concluded that the crown-down technique was superior to the step-back technique in the preparation of canal curvatures ranging from 10 to 35 degrees [12]. Regardless of the instrument or technique used, Weine concluded that prepared canals showed characteristics reflecting the inability to maintain the general shape of the curved canal [10]. Lim and Weber determined that stiffer files resulted in greater apical transportation and speculated that more flexible files might limit these undesirable outcomes [15].

William J. Buehler originally discovered Nitinol (*N*ickel-*T*itanium-*N*aval *O*rdnance *L*aboratory) in 1959 during his research at the Naval Ordnance Laboratory in White Oaks, Maryland [46]. Despite extensive use in orthodontics, Nitinol was not used in endodontics until 1988 when Walia et al. proposed the use of Nitinol nickel-titanium orthodontic wire to fabricate endodontic files [16]. Walia's landmark paper was the first to investigate new metallurgic properties to achieve better outcomes during root canal instrumentation. Citing superior flexibility due to its low modulus of elasticity, Walia et al. theorized that nickel-titanium root canal instruments would yield less procedural errors when instrumenting canal curvatures [16]. Using Nitinol orthodontic wires, fluted triangular cross-sectional shapes were machined directly onto size #15 wire blanks and evaluated based on cantilever bending, clockwise torsion and counterclockwise torsion as defined by Krupp et al. [16, 47]. Walia et al. concluded that Nitinol had a very low modulus of elasticity, one fourth to one fifth the value of stainless steel, resulting in superior elastic flexibility and resistance to fracture [16].

Nickel-titanium instruments are widely used, as they have proven to be far more flexible than traditional stainless steel instruments and substantially reduce the incidence of procedural errors such as transportation and ledging in curved canals [48]. Pettiette et al. found that nickel-titanium instruments produced significantly less canal transportation in the apical third of molars than stainless steel hand instruments in the hands of fourth year dental students [48]. Furthermore, Glosson et al. found that the more flexible nickel-titanium instruments used in engine-driven rotary handpieces had several advantages over hand instrumentation with traditional K-flex files including more centered canal preparations and less transportation [49].

Despite the increased flexibility, clinicians reported increased instrument separation during root canal instrumentation with nickel-titanium files [17, 50]. In 1995, Camps et al. concluded that nickel-titanium instruments actually presented lower torque values at failure than stainless steel instruments resulting in a higher incidence of instrument separation [17]. Pruett et al. found that unlike stainless steel instruments that show visible signs of deformation, nickel-titanium instruments can fracture without any visible defects of permanent deformation [50]. Consequently, a great deal of research was conducted in the early 1990s investigating nickel-titanium instruments and reasons for fracture. Researchers concluded that instruments used in a rotary motion fracture in two distinct ways - torsional fracture and cyclic flexural fatigue [51]. Torsional fracture refers to how much a file can rotate before its plastic limit is reached and the instrument fractures [51]. Cyclic fatigue is caused by repeated tensile and compressive stresses on a file in the area of the curvature [50-52].

In addition to instrument separation nickel-titanium instruments exhibited other shortcomings. Specifically, nickel-titanium instruments were shown to wear faster and have decreased cutting efficiency when compared to traditional stainless steel hand instruments [18, 53]. Coleman and Svec compared step-back preparations in curved canals of resin blocks using nickel-titanium K-files and stainless steel K-files. While their results showed that nickel-titanium instruments caused significantly less transportation than stainless instruments, nickel-titanium instruments took significantly longer to prepare resin blocks than stainless steel instruments [54].

In an effort to reduce torsional fracture and increase cutting efficiency, Tulsa Dental introduced a nickel-titanium file with greater taper, referred to as the 0.04 mm "U" file. According to the manufacturer, these files increased the cutting efficiency from previous generations while reducing instrument failure and reducing the risk of canal transportation [55]. While these changes resulted in more efficient instruments, the stiffness of the instruments increased resulting in more canal transportation towards the outer aspect of the curvature in the apical region of root canals [19, 20].

In 1997, Wolcott and Himel evaluated the torsional properties of stainless steel K-type .02 taper and nickel-titanium U-type .02 and .04 taper instruments using the current ANSI/ADA Specification #28 [55]. While the torsional properties of both instruments were within the acceptable tolerances as defined by ADA Specification #28, Wolcott and Himel stated that comparing the torque values for the nickel-titanium .04 instruments to the values set forth in ADA Specification #28 for .02 stainless steel instruments had its limitations [55]. Specifically, the taper of the nickel-titanium instruments (0.04mm) was twice that of conventional stainless steel instruments resulting in different tip sizes between the two groups at any given point along the file [55]. Walcott and Himel concluded that these differences might result in greater disparities in tolerances set forth by current testing standards [55].

In 1992, the International Standards Organization set forth new requirements and testing methods for root canal instruments known as ISO 3630-1. According to these standards, resistance to bending (i.e. flexibility) of root canal instruments (stainless steel or nickel-titanium) is measured by fixing the instrument 3mm from its tip and bending it. The bending moment is measured when the angular deflection reaches 45° [21]. This approach does not account for the variety of instruments currently available and the effect they have as canal curvature changes.

Since these standards were first adopted in 1992 research involving nickel-titanium instruments has risen dramatically. Newer generations of nickel-titanium instruments such as Brasseler’s (Brasseler USA Dental, Savannah, GA) EndoSequence system have been introduced. These files are made from conventional nickel-titanium triangular blanks without radial lands, and incorporates alternating contact points, which the manufacturer claims to enable the file to stay more centered within the canal thereby reducing canal transportation [56]. While these instruments have been shown to have superior flexibility from previous generations, the inability to maintain original canal anatomy during cleaning and shaping continues to be a problem [57-59].

MATERIALS AND METHODS

In this study, the flexibility of stainless steel hand files (Roydent Dental Products, Johnson City, TN) and EndoSequence nickel-titanium rotary files (Brassler USA Dental, Savannah, GA) of various tapers was measured. This study utilized ISO size 30 files as this apical preparation size has been shown to effectively reduce the bacterial count while maintaining the original canal anatomy [60].

Table 1 *– Files tested*

Brand	Type	Taper	Tip Size	Sample Size		
				3mm	*5mm*	*7mm*
Roydent	SS	0.02	30	10	10	10
Brasseler EndoSequence	NiTi	0.04	30	10	10	10
Brasseler Endosequence	NiTi	0.06	30	10	10	10

Each file was taken directly from the manufacturer's packaging and measured 25 mm from the tip (i.e. D_o) to the handle. The sample size was 10 for each type, taper and size in accordance with the instructions given in ISO 3630-1 [21]. The tips of each file were measured at 3 mm, 5 mm or 7 mm using a digital caliper and marked with a rubber stopper. A constant deflection point of 20 mm from D_0 was verified using a digital caliper. Each file was placed into a chuck and securely fastened between two metal plates on a load-sensing cell. Bending was accomplished using a universal testing machine (Instron Model 5500R, Norwood, MA) to a maximum deflection of 4.5 mm at a rate of

2mm/minute under room temperature conditions ($22^{O}C \pm 1°C$). Raw data was collected electronically via Merlin Software and transferred to Microsoft Excel for further analysis. Statistical analysis was completed with IBM SPSS Statistics software (IBM Corporation, Armonk, NY) and a two-way analysis of variance (ANOVA) was used as well as a post-hoc Tukey test.

Figure 1 *- Stainless steel #30 hand file*

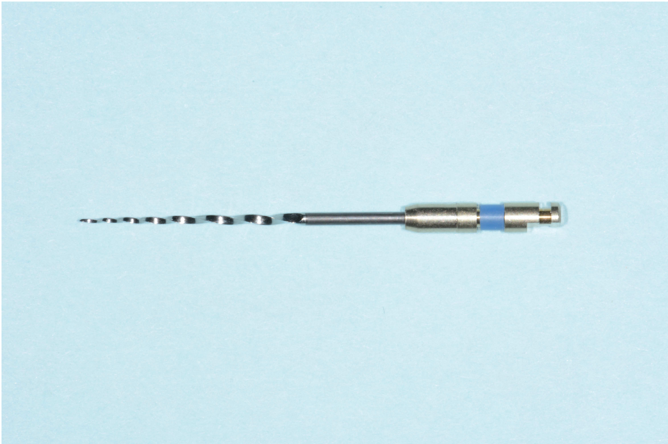

Figure 2 *- EndoSequence #30 .04 NiTi rotary file*

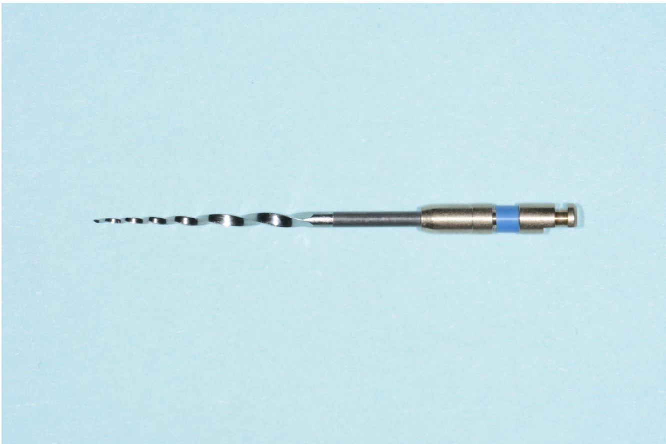

Figure 3 *- EndoSequence #30 .06 NiTi rotary file*

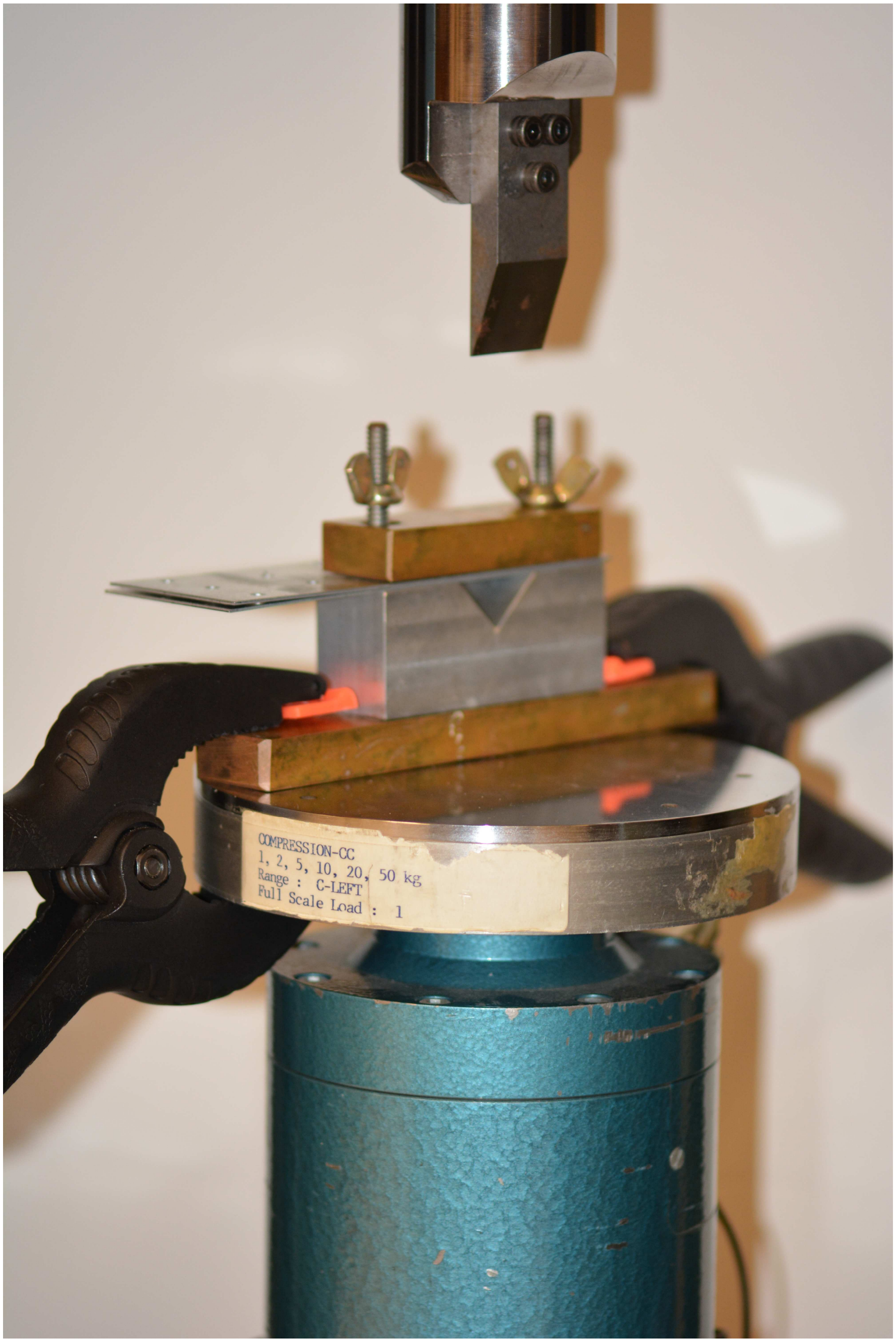

Figure 4 - *Testing apparatus*

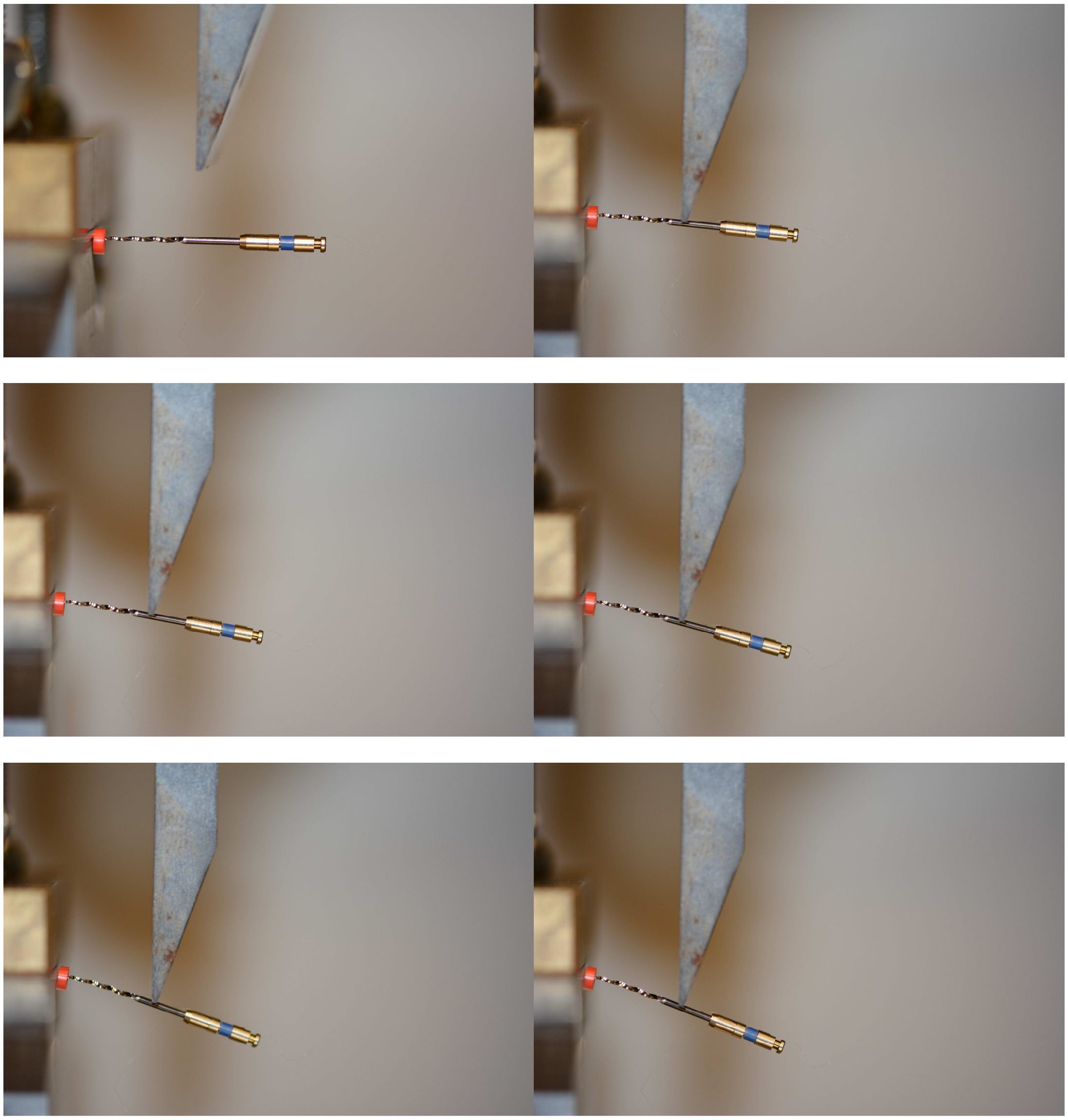

***Figure 5** - Deflection from 0 mm to 4.5 mm*

RESULTS

Data for stiffness (g/mm), average force (g) and bending moments (g*mm) are shown in Table's 2 and 3 below. Data was analyzed using a two-factor analysis of variance (ANOVA) and post-hoc Tukey test for both file and grasp length.

Grasp Length

The force and bending moments were significantly greater ($p<0.05$) within each file group (EndoSequence 0.06, EndoSequence 0.04 and Stainless steel 0.02) as the grasp length increased; 3 mm < 5 mm < 7 mm.

Force

The force (g) to bend EndoSequence 0.06 tapered files was significantly greater ($p<0.05$) than both EndoSequence 0.04 tapered files and stainless steel hand files from initial deflection to the maximum deflection of 4.5 mm. No significant differences were noted between EndoSequence 0.04 tapered files and stainless steel hand files from initial bending to 3.0 mm. From 3.5 mm to the maximum deflection of 4.5 mm, the force required to bend stainless steel hand files was significantly greater ($p< 0.05$) than EndoSequence 0.04 tapered files.

Bending Moment

The bending moment (g*mm) for EndoSequence 0.06 tapered files was significantly greater ($p<0.05$) than both EndoSequence 0.04 tapered files and stainless steel hand files from initial deflection to the maximum deflection of 4.5 mm. No significant differences were noted between EndoSequence 0.04 tapered files and stainless steel hand files from initial bending to 3.0 mm. From 3.5 mm to the maximum deflection of 4.5 mm, the bending moment for stainless steel hand files was significantly greater ($p < 0.05$) than EndoSequence 0.04 tapered files. See figures 6-14 below for graphical analysis of the force vs. deflection curves as well as the bending moment vs. deflection curves for Stainless steel, EndoSequence 0.04 and EndoSequence 0.06 files.

Table 2 *– Force exhibited by the files at various deflections*

File	Grasph Length	Stiffness (g/mm)	Force (g)										
			0.25mm	0.50mm	0.75mm	1.0mm	1.5mm	2.0mm	2.5mm	3.0mm	3.5mm	4.0mm	4.5mm
Stainless steel	3mm	3.73 ± 0.51	1.01 ± 0.18	2.03 ± 0.32	3.02 ± 0.43	3.93 ± 0.54	5.79 ± 0.88	7.50 ± 1.14	9.15 ± 1.51	10.70 ± 1.78	12.11 ± 1.98	13.47 ± 2.18	14.86 ± 2.55
Sequence 0.04	3mm	3.36 ± 0.71	1.10 ± 0.21	2.03 ± 0.34	2.91 ± 0.53	3.75 ± 0.66	5.33 ± 1.03	6.83 ± 1.26	8.17 ± 1.49	9.40 ± 1.57	10.43 ± 1.62	11.18 ± 1.48	12.02 ± 1.62
Sequence 0.06	3mm	7.39 ± 1.47	2.15 ± 0.35	4.14 ± 0.69	6.11 ± 1.06	7.89 ± 1.39	11.16 ± 2.06	14.12 ± 2.64	16.76 ± 3.23	18.91 ± 3.56	20.58 ± 3.83	21.38 ± 4.13	21.80 ± 4.10
Stainless steel	5mm	5.46 ± 0.60	1.40 ± 0.17	2.78 ± 0.33	4.20 ± 0.44	5.51 ± 0.58	8.08 ± 0.82	10.58 ±1.03	12.88 ±1.22	15.11 ±1.38	16.99 ±1.65	18.82 ± 1.71	20.68 ± 2.00
Sequence 0.04	5mm	5.61 ± 1.25	1.47 ± 0.57	2.91 ± 0.90	4.32 ± 1.23	5.71 ± 1.51	8.13 ± 1.95	10.25 ± 2.44	12.24 ±2.57	14.06 ± 2.81	15.57 ± 2.73	16.56 ± 2.82	17.35 ± 2.65
Sequence 0.06	5mm	12.81 ± 3.24	3.96 ± 0.90	7.45 ± 1.79	10.84 ± 2.66	13.90 ±3.41	19.66 ± 4.47	24.71 ± 5.43	29.35 ± 6.11	32.74 ± 6.71	35.18 ± 6.77	36.69 ± 6.65	38.02 ± 6.41
Stainless steel	7mm	13.97 ± 0.92	3.65 ± 0.32	7.11 ± 0.53	10.68 ± 0.79	14.14 ± 0.97	20.62 ± 1.49	26.56 ± 1.93	32.22 ± 2.32	37.38 ± 2.73	41.84 ± 3.19	45.55 ±3.56	49.08 ± 4.00
Sequence 0.04	7mm	13.12 ± 2.72	3.97 ± 0.77	7.56 ± 1.50	11.02 ± 2.25	14.18 ± 2.84	19.91 ± 3.58	24.86 ± 4.09	28.99 ± 4.24	31.94 ± 4.19	33.93 ± 3.90	34.82 ± 3.79	35.45 ± 3.95
Sequence 0.06	7mm	40.74 ± 6.09	12.36 ± 1.45	23.66 ± 3.14	34.63 ± 4.79	44.10 ± 6.12	61.18 ± 8.28	74.91 ± 10.37	84.66 ± 11.42	89.75 ± 11.77	92.99 ± 12.27	95.93 ± 12.39	98.93 ± 12.63

Table 3 *- Bending moment exhibited by the files at various deflections*

File	Grasp Length	Stiffness (g/mm)	Bending Moment (g*mm)										
			0.25mm	0.50mm	0.75mm	1.0mm	1.5mm	2.0mm	2.5mm	3.0mm	3.5mm	4.0mm	4.5mm
Stainless steel	3mm	3.73 ± 0.51	17.22 ± 3.11	34.59 ± 5.36	51.27 ± 7.28	66.77 ± 9.23	98.47 ± 14.92	127.49 ± 19.33	155.59 ± 25.62	181.98 ± 30.24	205.80 ± 33.68	228.98 ± 37.04	252.62 ± 43.30
Sequence 0.04	3mm	3.36 ± 0.71	18.76 ± 3.49	34.56 ± 5.80	49.52 ± 9.06	63.77 ± 11.24	90.62 ± 17.45	116.17 ± 21.35	138.90 ± 25.35	159.82 ± 26.61	177.33 ± 27.55	190.00 ± 25.13	204.42 ± 27.61
Sequence 0.06	3mm	7.39 ± 1.47	36.63 ± 5.89	70.43 ± 11.69	103.92 ± 18.08	134.09 ± 23.57	189.76 ± 35.08	240.10 ± 44.85	284.92 ± 54.92	321.45 ± 60.54	349.91 ± 65.12	363.49 ± 70.25	370.59 ± 69.66
Stainless steel	5mm	5.46 ± 0.60	21.05 ± 2.58	41.70 ± 4.88	63.04 ± 6.61	82.59 ± 8.74	121.24 ± 12.29	158.64 ± 15.39	193.17 ± 18.29	226.67 ± 20.66	254. 89 ± 24.70	282.26 ± 25.69	310.16 ± 30.06
Sequence 0.04	5mm	5.61 ± 1.25	22.07 ± 8.60	43.62 ± 13.43	64.79 ± 18.49	85.58 ± 22.66	121.92 ± 29.27	153.74 ± 36.66	183.53 ± 38.54	210.92 ± 42.12	233.53 ± 40.89	248.45 ± 42.23	260.24 ± 39.75
Sequence 0.06	5mm	12.81 ± 3.24	59.34 ± 13.52	111.80 ± 26.82	162.53 ± 39.89	208.48 ± 51.13	294.90 ± 66.99	370.58 ± 81.52	440.26 ± 91.63	491.16 ± 100.72	527.77 ± 101.62	550.33 ± 99.72	570.37 ± 96.12
Stainless steel	7mm	13.97 ± 0.92	47.39 ± 4.10	92.41 ± 6.90	138.87 ± 10.26	183.83 ± 12.61	268.06 ± 19.38	345.27 ± 25.15	418.83 ± 30.19	486.00 ± 35.46	543.89 ± 41.51	592.10 ± 46.23	638.09 ± 52.05
Sequence 0.04	7mm	13.12 ± 2.72	51.67 ± 10.05	98.32 ± 19.49	143.23 ± 29.22	184.31 ± 36.94	258.82 ± 46.58	323.12 ± 53.20	376.84 ± 55.16	415.25 ± 54.45	441.08 ± 50.65	452.69 ± 49.22	460.85 ± 51.37
Sequence 0.06	7mm	40.74 ± 6.09	160.73 ± 18.81	307.64 ± 40.88	450.21 ± 62.32	573.30 ± 79.52	795.34 ± 107.68	973.85 ± 134.84	1100.55 ± 148.47	1166.81 ± 152.95	1208.93 ± 159.48	1247.07 ± 161.11	1286.09 ± 164.16

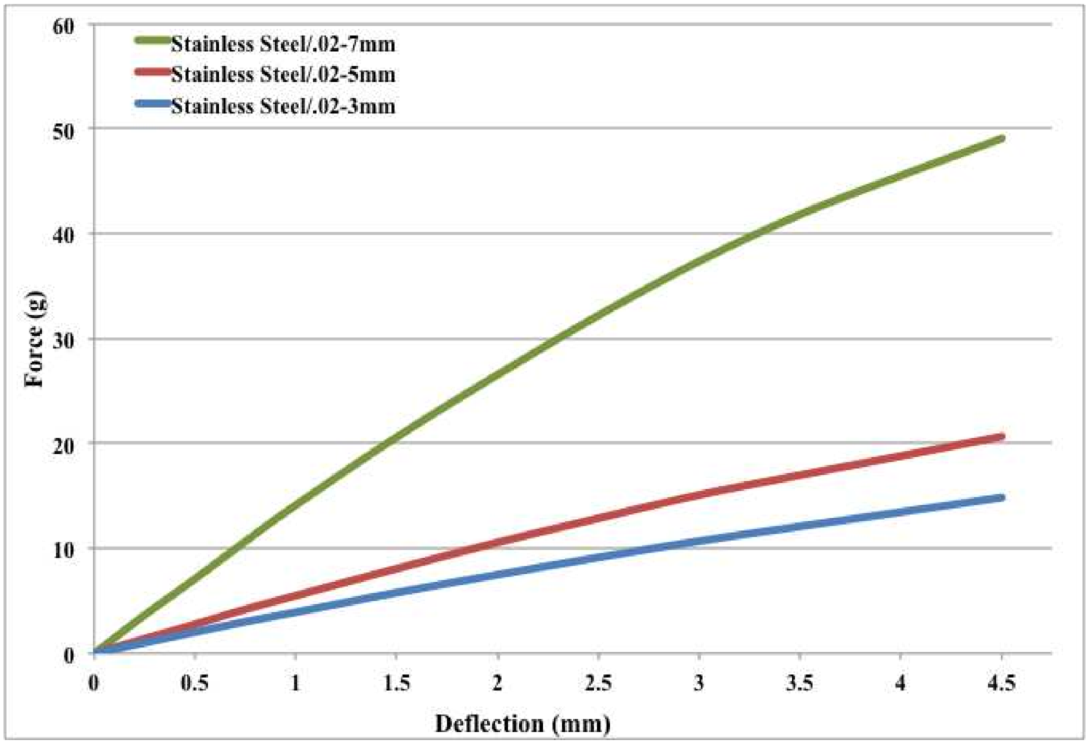

***Figure 6** – Force vs. deflection for stainless steel files*

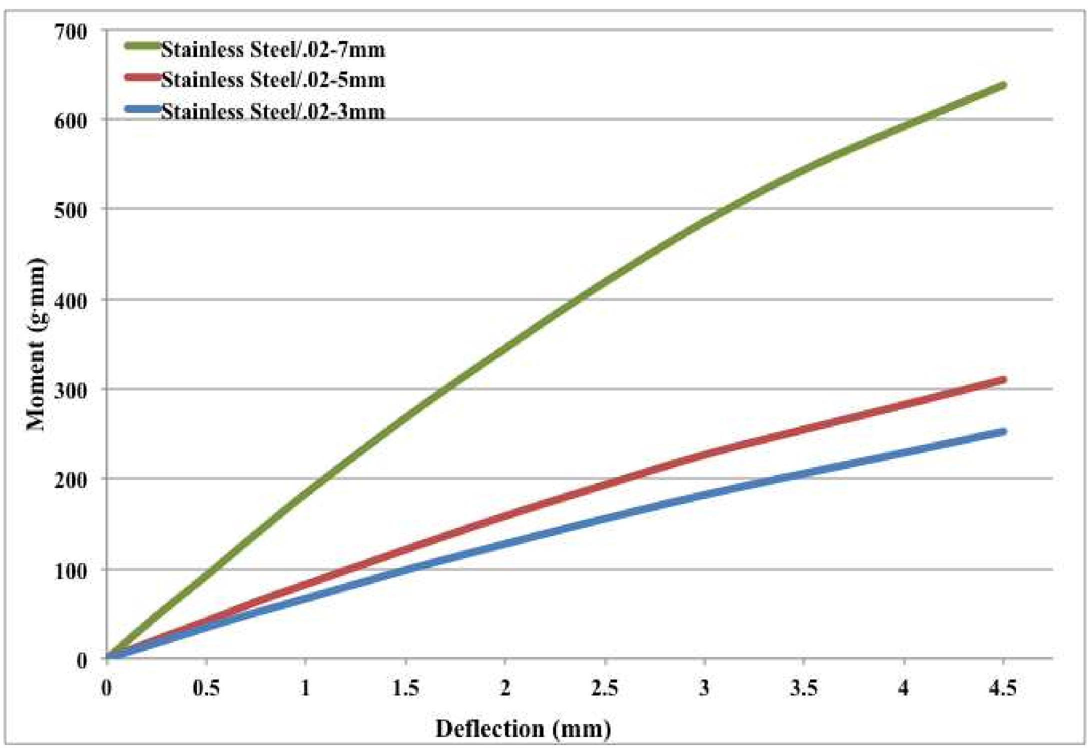

***Figure 7** – Bending moments vs. deflection for stainless steel files*

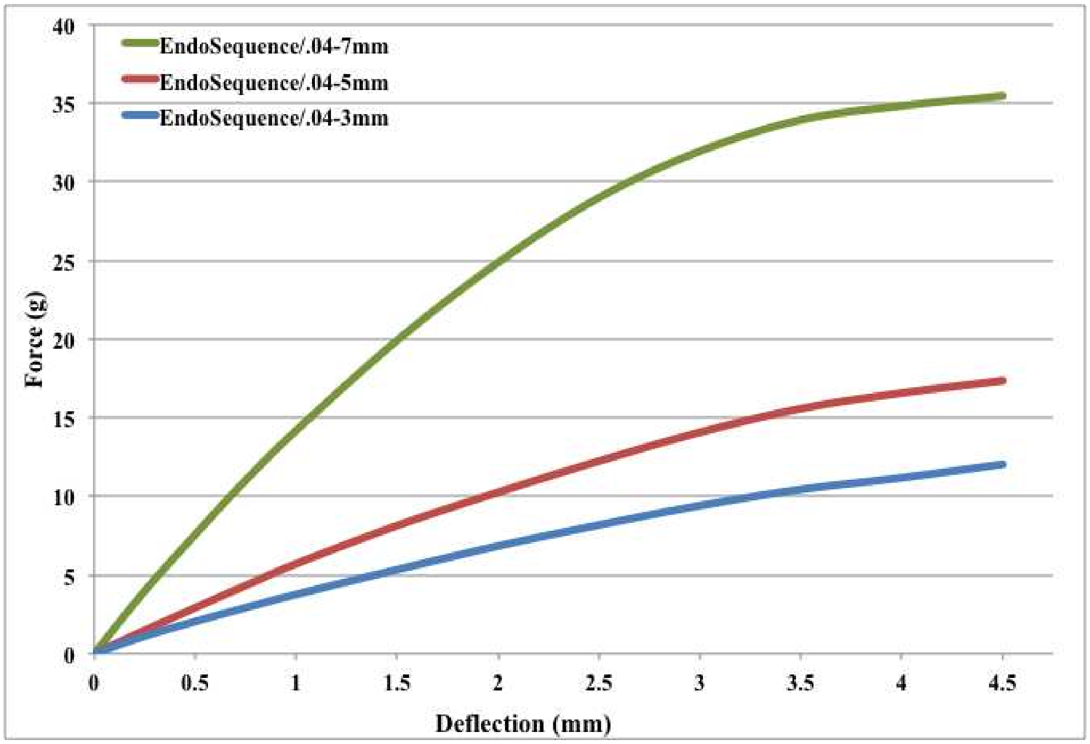

Figure 8 *– Force vs. deflection EndoSequence 0.04 files*

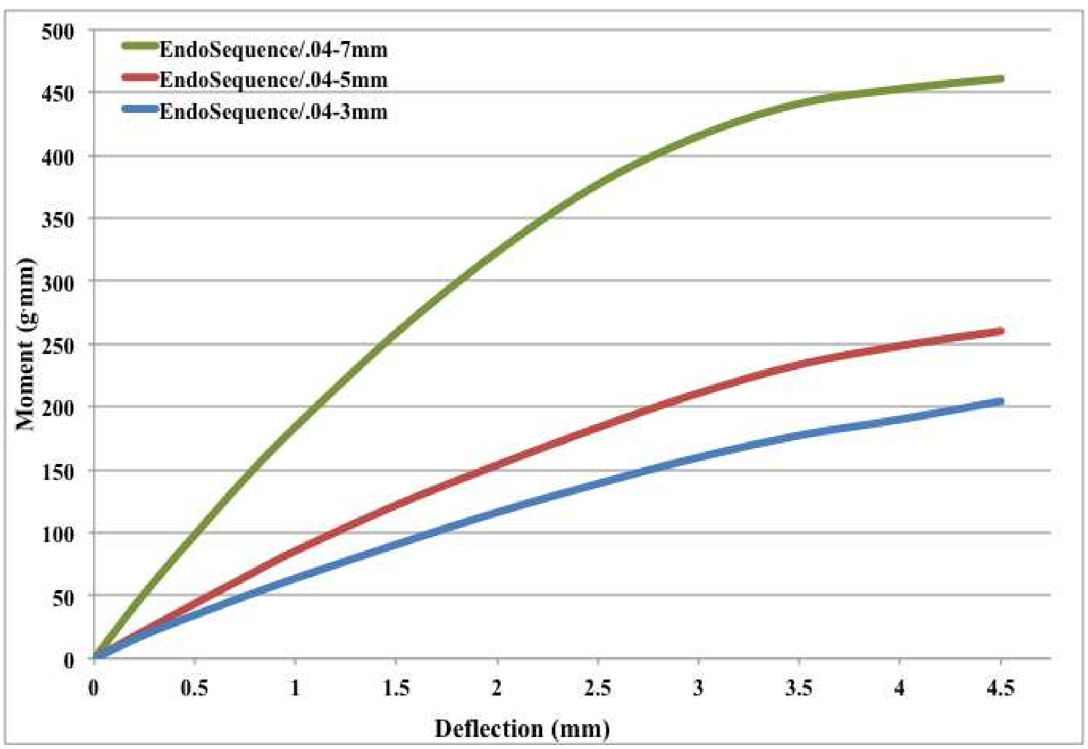

Figure 9 *– Bending moment vs. deflection EndoSequence 0.04 files*

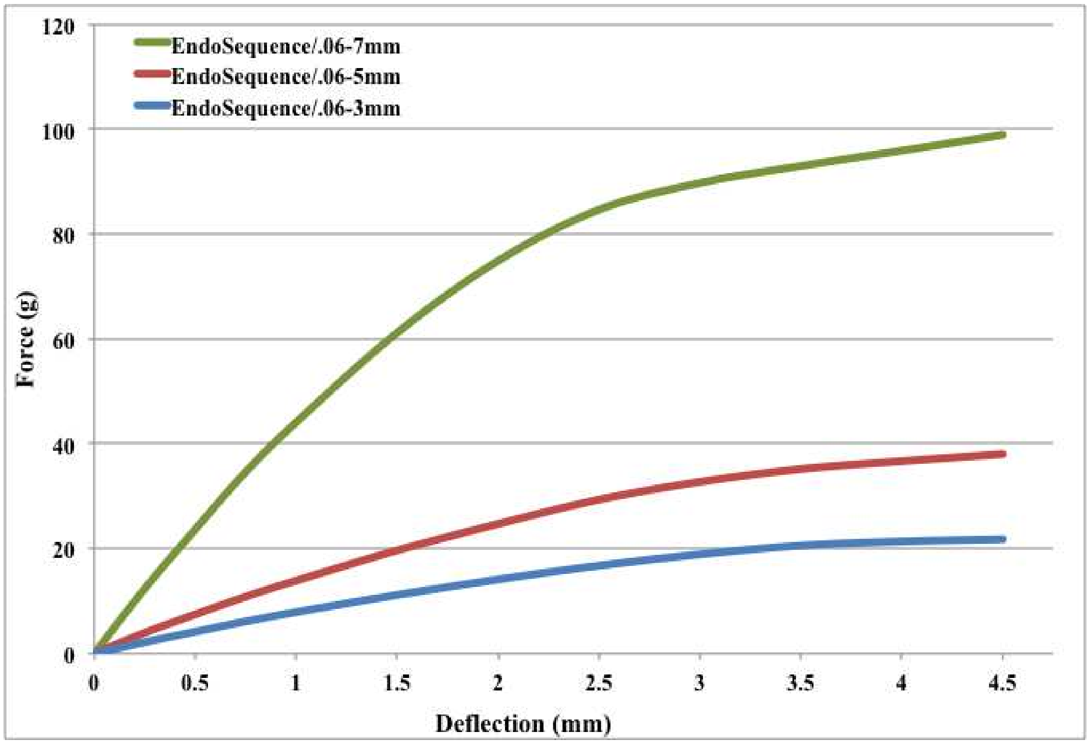

Figure 10 *- Force vs. deflection EndoSequence 0.06 files*

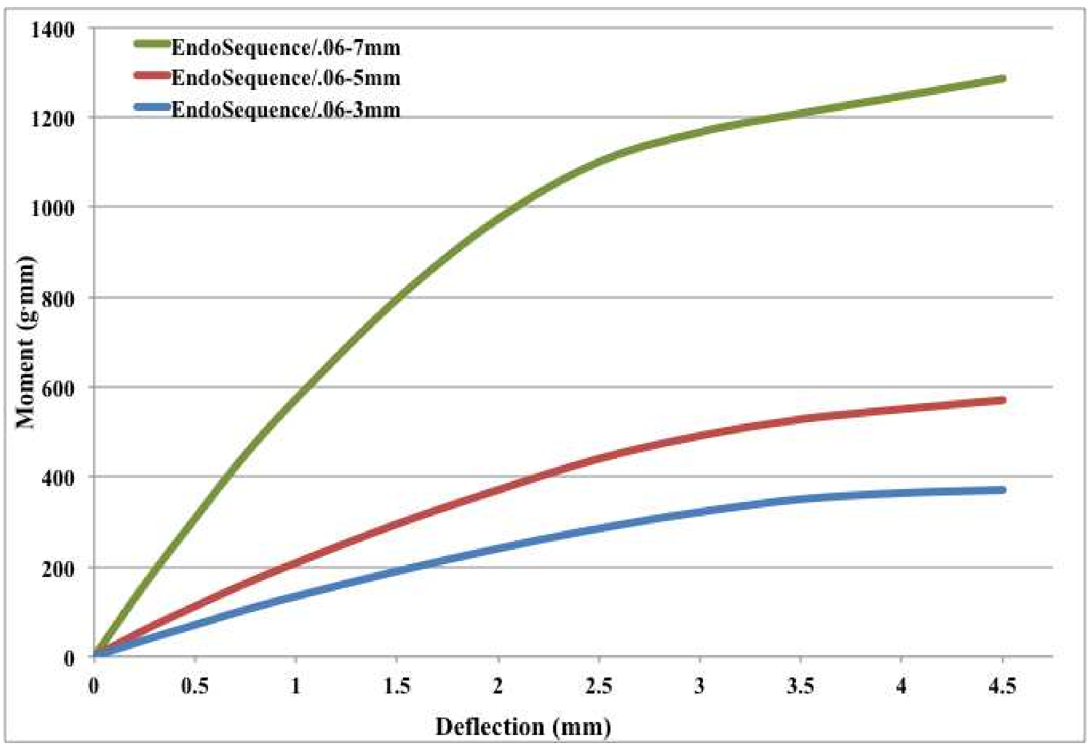

Figure 11 *- Bending moment vs. deflection EndoSequence 0.06 files*

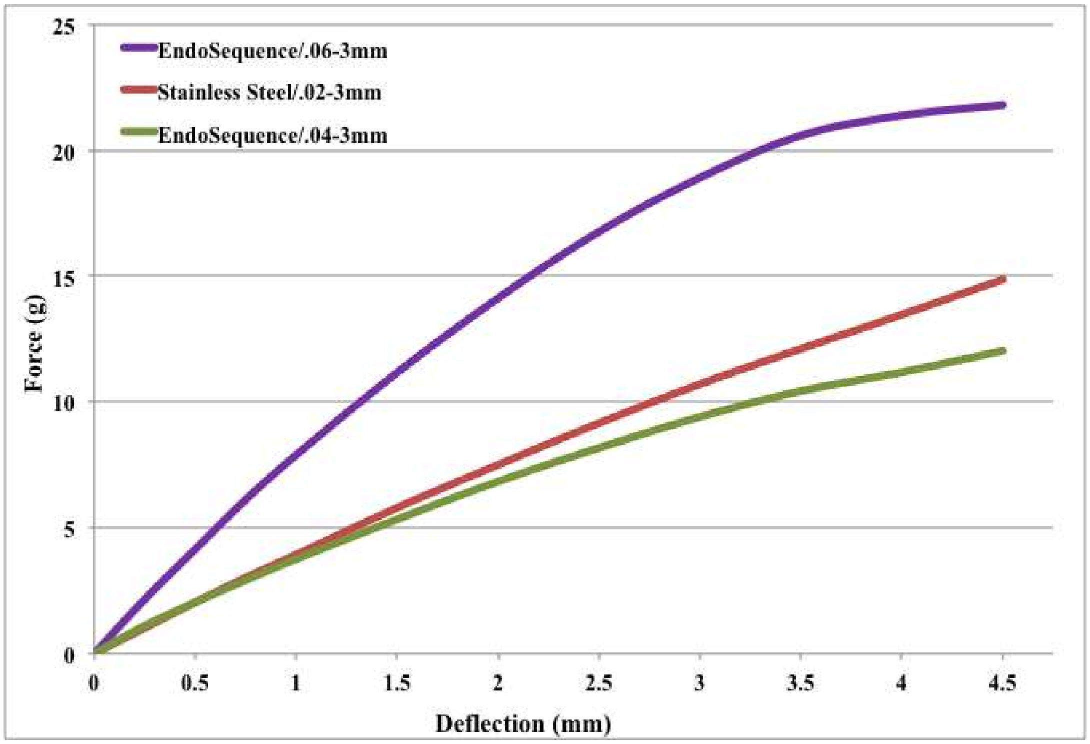

***Figure 12** – Force vs. deflection comparison of the files when grasped at 3 mm*

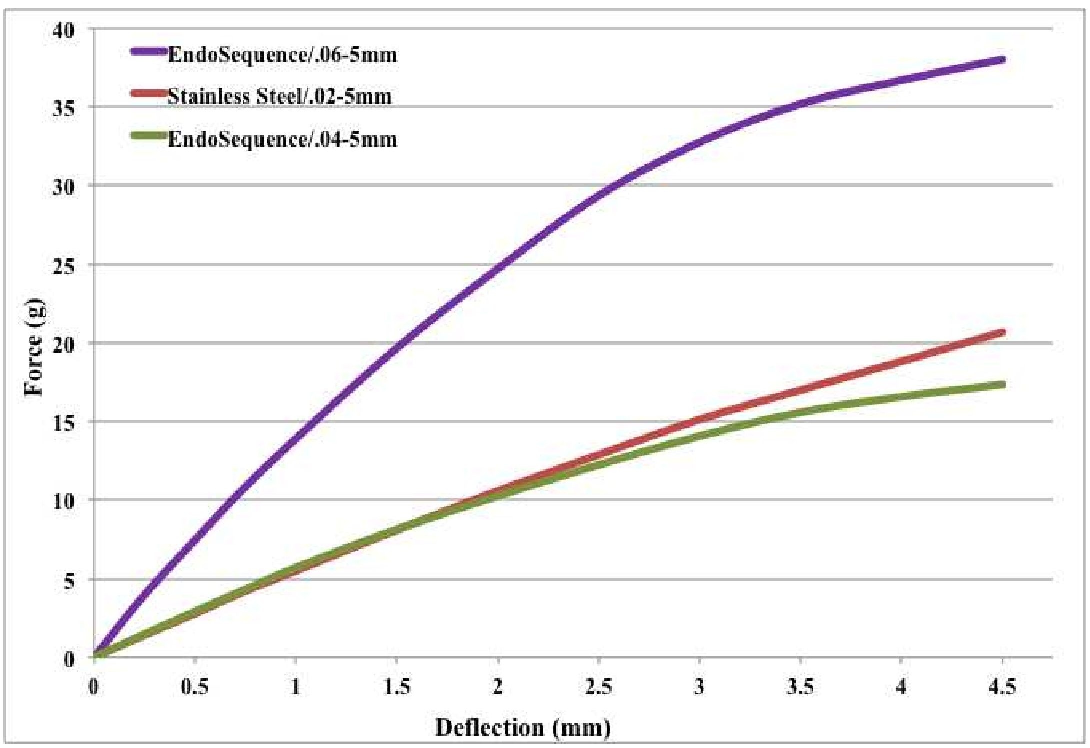

***Figure 13** - Force vs. deflection comparison of the files when grasped at 5 mm*

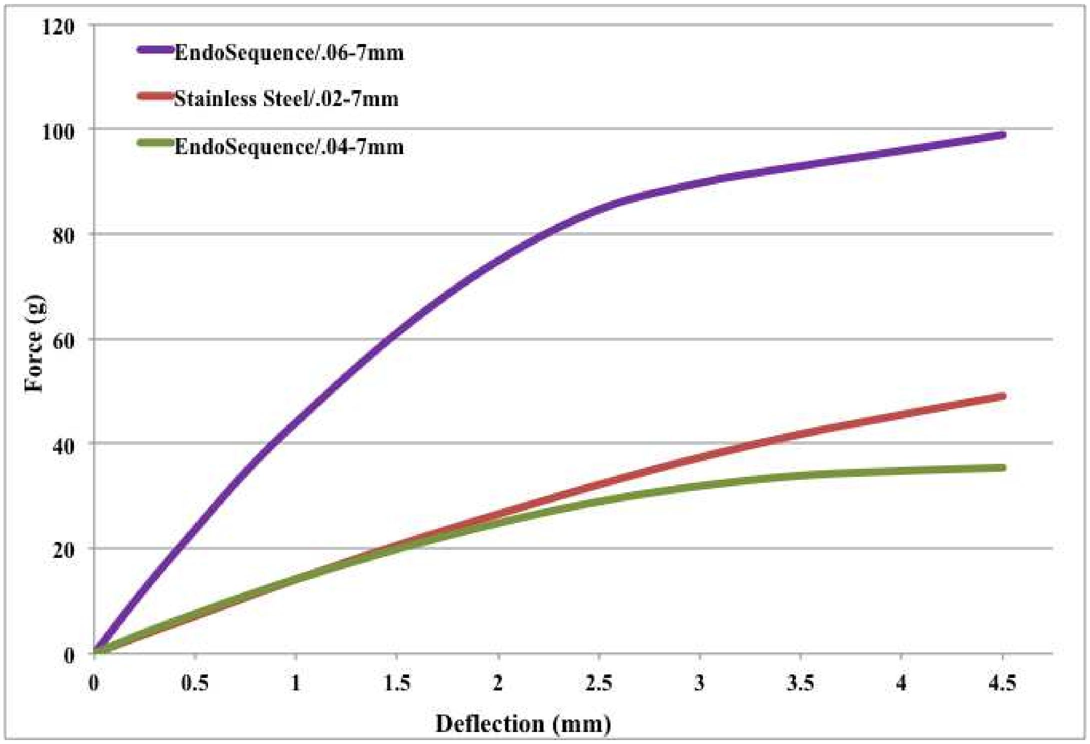

Figure 14 - *Force vs. deflection comparison of the files when grasped at 7 mm*

DISCUSSION

The fundamental objective of non-surgical root canal therapy is to prevent or resolve pulpal pathology and periapical pathosis. [3] Proper cleaning and shaping should result in elimination of microorganisms from the canal system while simultaneously providing a continuously tapered canal to facilitate obturation. [1] Cleaning and shaping is a chemo-mechanical process in which mechanical instrumentation plays a crucial role by removing infected dentin and facilitating the apical flow of irrigants to disinfect the canal system. Complex canal anatomy provides challenges for clinicians during mechanical instrumentation that may prevent adequate disinfection of the root canal system. [50] Complications such as transportation, perforation, ledging and instrument separation often occur in the presence of complex canal anatomy.[8-10] Prior to the introduction of nickel-titanium instruments, several instrumentation techniques were developed to overcome these challenges. While these techniques were widely accepted, Weine proved that all techniques showed an inability to maintain the general shape of the original canal anatomy. [10]

In 1988, Walia et al. introduced nickel-titanium instruments, which had three times the flexibility of traditional stainless steel hand instruments. [16] Numerous studies have shown that the use of nickel-titanium instruments minimizes procedural errors resulting in more predictable outcomes. [50] While nickel-titanium alloys exhibit improved elasticity; this elasticity is limited by the size and taper of the instrument used. [61] Instruments of greater taper have been shown to have increased stiffness resulting in an increased risk of transportation in curved canals. [61]

Current ISO standards specify maximum values for stiffness and bending moments for root canal instruments with 0.02 mm taper. While this approach is widely accepted, these standards do not account for instruments with greater taper and the effect they have on the file's flexibility. The purpose of this study was to measure the bending moment of root canal instruments of different tapers at three different points along each file to a maximum deflection of 4.5 mm to determine the forces required to bend the instruments.

The results of this study indicate that the degree of taper affects the amount of force required to bend files. The stiffness test revealed that the bending moments for the EndoSequence 0.06 tapered instruments were significantly greater at all lengths than the bending moments for EndoSequence 0.04 tapered instruments and the stainless steel hand files with 0.02 taper. These results correspond with previous findings that greater tapered instruments (> 0.04) are significantly stiffer than those of lesser-tapered instruments. [33, 50] No significant differences were noted between EndoSequence 0.04 tapered files and stainless steel hand files from initial bending to 3.0 mm. However, from 3.5 mm to maximum deflection of 4.5 mm, the bending moment for stainless steel hand files was significantly greater ($p< 0.05$) than EndoSequence 0.04 tapered files. These results agree with Walia et al.'s original findings that nickel-titanium instruments are more flexible than traditional stainless steel hand instruments and are able to negotiate canal curvatures more easily resulting in less apical transportation. [16]

Surprisingly, little research exists regarding the bending properties of root canal instruments at different grasp lengths. A review of the literature showed that most research follows ISO 3630-1 and measures the bending moment by grasping the file 3mm from the tip and bending the file 45°. The results of this study showed that in addition to taper, grasp length also greatly influences the amount of force required to bend files. Specifically, the force and bending moments were significantly greater ($p<0.05$) within each file group (EndoSequence 0.06, EndoSequence 0.04 and Stainless steel 0.02) as the grasp length increased; 3 mm < 5 mm < 7 mm. As root canal anatomy is diverse and no two canals are similar new testing protocols that account for canal curvatures at various points throughout the canal should be considered.

CONCLUSION

During root canal preparation, endodontic instruments are subjected to a variety of forces as canal anatomy changes. Therefore instruments should have properties capable of minimizing the possibility of undesirable procedural errors such as canal transportation and ledging. Although nickel-titanium instruments were introduced to eliminate undesirable changes, studies have shown that as stiffness increased the incidence of transportation in the apical region of canals increased as well. The results of the present study confirmed previous findings and indicate that nickel-titanium instruments of greater taper significantly affects the amount of force required to bend a file thereby decreasing the file's flexibility. Furthermore, regardless of the type of instrument used (i.e. stainless steel or nickel-titanium) the point of deflection on the file significantly affects the force required to bend the file.

With a vast array of root canal instruments currently available, clinicians should consider the metallurgical properties of instruments before cleaning and shaping. Specifically, nickel-titanium files with tapers greater than 0.04 mm should not be used for apical enlargement of curved canals because these files are significantly stiffer than 0.02 mm and 0.04 mm files resulting in an increased likelihood of canal transportation.

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