Original Article

Forces in initial archwires during leveling and aligning: An *in-vitro* study

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Abstract

Aim: This *in-vitro* study measured the force deflection behavior of selected initial alignment archwires by conducting three-point bending tests under controlled tests. The study tested three wire designs, namely, co-axial multistranded stainless steel wires, nickel–titanium, and copper–nickel–titanium archwires. **Materials and Methods:** The archwires were ligated to a specially designed metal jig, simulating the arch. A testing machine (Instron) recorded activation and deactivation forces of different deflections at 37°C. Forces on activation and deactivation were compared by one-way analysis of variance (ANOVA). **Results:** Significant differences ($P < 0.05$) in activation and deactivation forces were observed among the tested wires. The co-axial multistranded wire had the lowest mean activation and deactivation forces, whereas conventional nickel–titanium wires had more mean activation and deactivation forces at different deflections. **Conclusion:** The activation and deactivation forces were higher for nickel–titanium followed by copper–nickel titanium and co-axial wires. The amount of percentage force loss was more for co-axial wire, indicating that these wires are not ideal for initial leveling and aligning.

Key words: *Activation force, deactivation force, leveling and aligning, stiffness*

INTRODUCTION

The first stage of orthodontic treatment entails leveling and aligning. During this stage, archwires with desirable stiffness are required to correct vertical and horizontal discrepancies. The forces in play during leveling and aligning are deactivation forces, and hence clinicians should have knowledge of deactivation forces to level and align the malpositioned teeth.[1]

In early 1930s, the only orthodontic wires available were made of gold. Later, austenitic stainless steel wires, with greater strength, high modulus of elasticity, good resistance to corrosion, and moderate cost were introduced as orthodontic wires in 1929, and shortly thereafter gained popularity over gold.[1]

Before the advent of contemporary alloys, various methods were used to maximize the desirable properties

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of stainless steel. One method was to bend loops into archwires to reduce the stiffness, thereby increasing the inter bracket distance. This posed problem of hygiene and tissue impingement, as well as increased chair-side time.[2]

Later, several contemporary alloys were developed and adopted in orthodontics, to mention a few, cobalt chromium, nickel–titanium, copper–nickel titanium and beta titanium.[3,4]

The first stage of orthodontic treatment entails leveling and aligning of the teeth. Usually, the ideal archwires for this vital stage generate a continuous and light force over a long period of time.[1,3,4]

The type of archwire most frequently recommended in contemporary practice for the initial stages of the orthodontic treatment is the super elastic nickel–titanium and thermo-activated type. However, some prefer multistranded steel archwires. These are cheaper and have not been shown to be less clinically effective than nickel–titanium wires. Laboratory tests can be helpful in the assessment of aligning archwires by providing information regarding the amount of forces exerted by the archwires on to the teeth. To accomplish this, an appliance that delivers forces that are light, decrease moderately between appointments, and with less discomfort to the patient is required.[5-7]

When a clinician engages a wire into the bracket slots of an appliance, energy is stored which represents activation forces. While the archwire tries to return to its original position, work is done on the dentition, as evidenced by the tooth movement which constitutes the deactivation forces.[8,9]

Hence, the force in play to level and align the teeth is not the activation force but the deactivation force or unloading force of the appliance. The activation and deactivation behaviors of a wire might not be the same. Therefore, force-deflection graphs generated during the activation (loading) and deactivation (unloading) cycles are not the same. Knowledge of the deactivation behavior is important to the clinician for optimal wire selection.[10,11]

The selection of an appropriate wire size, alloy type in turn, would provide the benefit of optimum and predictable treatment results.[12]

Therefore, the present study aims at evaluating and comparing the activation and deactivation forces in the initial archwires of nickel–titanium, copper–nickel titanium, and co-axial wires.

MATERIALS AND METHODS

The wires were selected as per the inclusion criteria of this study and were tested for their activation and deactivation forces. The methodology adopted for testing these specimens was the modified three-point bending test (Modified ADA Specification No. 32).

The archwires included in the present study were selected based on following criteria:

a. Absence of any surface defects on visual inspection.
b. Untampered sealed package.
c. Wires with recent manufactured date.

The sample size was calculated using the Z-score table and was calculated according to the equation

$$\text{Sample size} = (\text{Z-score})^2 \times \text{SD} \times (1 - \text{SD})/(\text{margin of error})^2$$

From the formula, a sample of 96 was calculated.

The samples comprised nickel–titanium, copper–nickel–titanium, and co-axial wires (32 each) and were randomized into six groups depending on the alloy content and size of the wire.

The samples were grouped as follows [Figure 1]:

Group I: Comprising 16 nickel–titanium archwires of 0.014-inch diameter.
Group II: Comprising 16 copper–nickel–titanium arch wires of 0.014-inch diameter.
Group III: Comprising 16 co-axial wires of 0.014-inch diameter.
Group IV: Comprising 16 nickel–titanium wires of 0.016-inch diameter.
Group V: Comprising 16 copper–nickel–titanium wires of 0.016-inch diameter.
Group VI: Comprising 16 co-axial wires of 0.016-inch diameter.

Metal jig

A specially designed cast iron jig was used for the study. The jig was U-shaped, simulating the arch form. A groove was made at the center along the outer surface of the jig for stabilizing or lodging the archwire. Three screws were placed to secure the archwire in the groove on either side of the jig. First pair of screws were placed at a distance of 9 mm in the anterior portion of the jig to simulate the average interbracket distance of the maxillary central incisors. Second and third pair of screws was placed at canine and first molar regions, respectively.

To facilitate the testing of activation and deactivation forces at various deflections, a portion of the jig was cut out in the anterior region measuring 6 mm in depth and 9 mm unsupported mid-span width between the first pair of screws [Figure 2].

Artificial saliva

Artificial saliva was used to simulate the oral environment. Composition used is mentioned in Table 1.

Brass vessel

A custom made brass vessel was used in this study to maintain the temperature of the artificial saliva during the experimental procedures. The temperature was monitored with the help of a thermometer.

The custom made brass vessel suited to the specially designed jig, and the base of the vessel was modified to fit the lower member of the Instron machine [Figure 2].

Instron machine with specially designed stylus

A stylus with the notch at the center to prevent the lateral deflection of the wire during the process of loading with a cross-head speed of 10 mm/min was specially designed [Figure 3].

Archwires were placed in the central groove on the outer surface of the jig and were secured with the screws. Later, the whole set up was placed in the brass vessel filled with artificial saliva maintained at 37°C, which represents the oral temperature as well as the transitional temperature range (TTR) of type III copper–nickel titanium. The temperature of the saliva was constantly monitored using a thermometer [Figure 2]. All the specimens were tested in artificial saliva, thereby simulating the oral conditions at its best.

Table 1: Composition of the artificial saliva

Inorganic constituents	Concentration mg/ltr
Ammonium chloride	233 mg/ltr
Calcium chloride dehydrate	210 mg/ltr
Magnesium chloride hexahydrate	43 mg/ltr
Potassium chloride	1162mg/ltr
Potassium dihydrogen orthophospate	354mg/ltr
Potassium thiocyanate	222mg/ltr
Sodium citrate	13mg/ltr
Sodium hydrogen carbonate	535 mg/ltr
Disodium hydrogen orthophosphate	375 mg/ltr
Aqua dislilta q.s.for	1000 ml
P^H	6.8

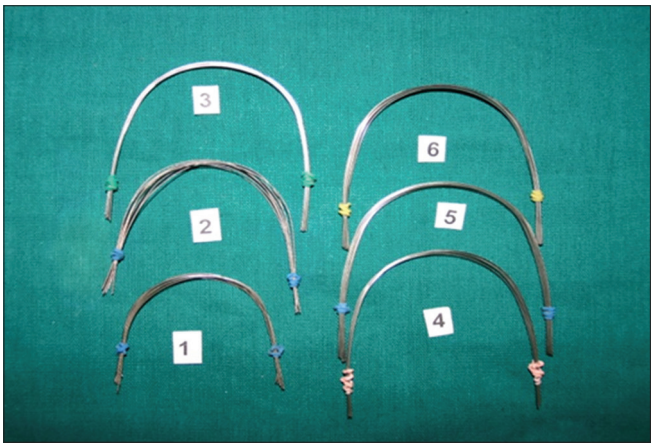


Figure 1: (1) 0.014 co-axial wires, (2) 0.016 co-axial wires, (3) 0.014 nickel titanium, (4) 0.016 nickel titanium, (5) 0.014 copper nickel titanium, (6) 0.016 copper nickel titan

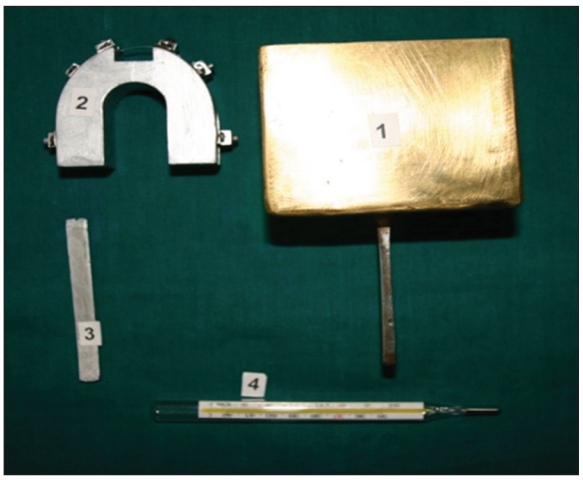


Figure 2: (1) Brass vessel, (2) Metal jig, (3) stylus and (4) thermometer

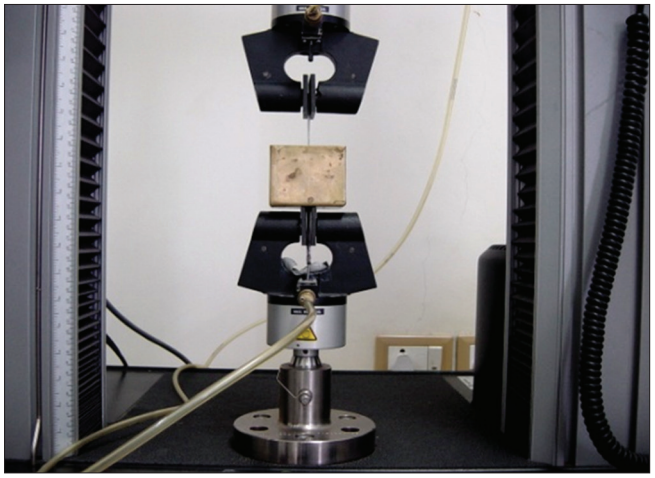


Figure 3: Specially fabricated jig mounted on the Instron Universal Testing Machine

Then, the brass vessel was mounted onto the lower member of the Instron Universal Testing Machine (Model#4467, Instron Corporation, Canton Mass). The specially designed stylus was connected to the cross-head of an Instron Universal Testing Machine, centered at the midspan of the wire specimen [Figure 4]. Then, the deflections of 0.5 mm, 1.0 mm, 1.5 mm, and 2.0 mm were produced with the stylus and the corresponding force values were tabulated as activation forces for that particular wire sample. Later, the wire samples were deflected to 2.1 mm and then deactivated to ensure that the deflections of −2.0 mm, −1.5 mm, −1.0 mm, and −0.5 mm were accurately produced and corresponding values were tabulated as deactivation forces. The whole procedure was done with 10-kg load cell and at the cross-head speed of 10 mm/min. The data were subjected to analysis of variance (ANOVA). Then, the percentage of force loss was calculated using the following formula:

Mean activation force − Mean deactivation force/Mean activation force × 100.

RESULTS

The results were analyzed using the one-way ANOVA, and the amount of deflection for each wire was calculated and compared with each other and tabulated. Table 2 shows the mean activation and deactivation forces (g) at different deflections for all the tested archwires.

From Graph 1, it is clearly evident that the loss of force among the 0.014-inch diameter wires was highest for the co-axial wires at all deflections, but it was very high when the deflection was 0.5 mm. The loss of force was equal for the nickel–titanium and copper–nickel–titanium at 1.5 mm deflection.

Graph 2 shows percentage loss of force among 0.016-inch diameter wires. The loss of force was the highest for 0.016 co-axial wires, and the loss of force was very much evident at all deflections. The nickel–titanium and copper–nickel–titanium showed loss of force at different deflections.

DISCUSSION

The strength, stiffness, and range of a wire are important to carry out a specific function at different stages of treatment.[1,5,12]

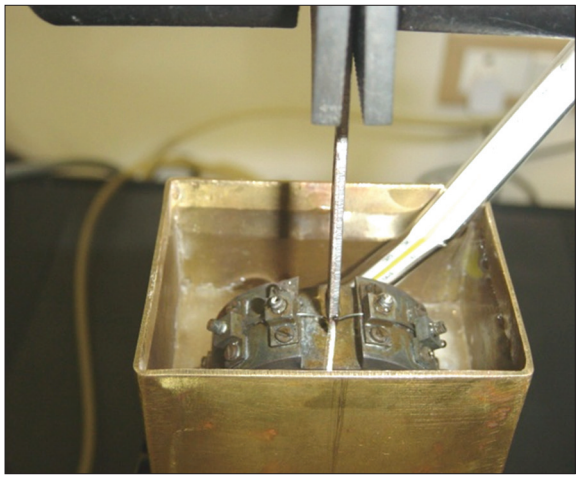
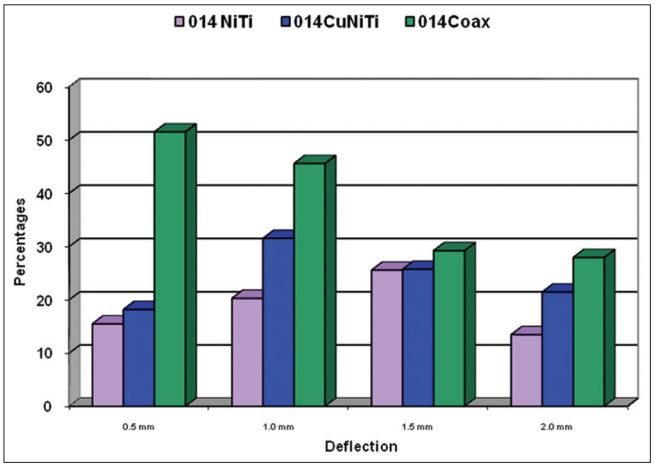
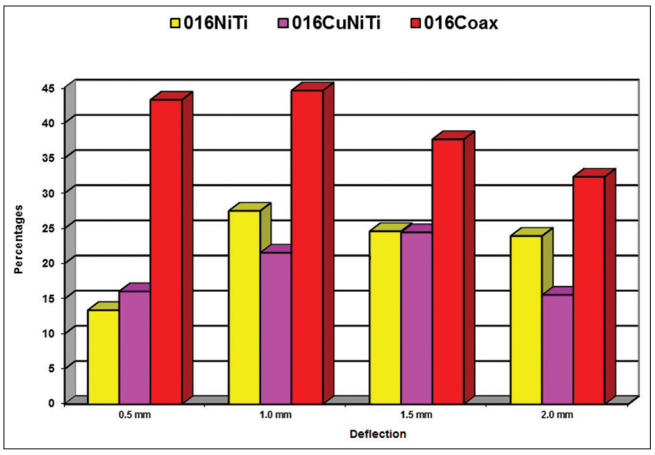


Figure 4: Mounted jig with the arch wire during testing



Graph 1: Percentage loss of force among 0.014 inch diameter wires at different deflections



Graph 2: Percentage loss of force among 0.016 inch diameter wires at different deflections

During the initial stage of the treatment, where initial leveling and alignment are desired, great range and light forces are sought. Ideally, archwires are designed to move teeth with light, continuous forces. According

Table 2: Mean activation and deactivation forces (gms) at different deflections for the tested arch wires

0.5 mm deflection		1.0 mm deflection		1.5 mm deflection		2.0 mm deflection	
Activation	Deactivation	Activation	Deactivation	Activation	Deactivation	Activation	Deactivation
80.4±13.4	67.9±2.1	149.9±5.6	119.5±3.6	219.2±10.3	163.2±3.9	255.1±12.1	220.8±11.6
70.4±2.1	57.6±2.2	137.1±3.8	93.9±2.7	212.6±4.9	157.9±4.5	231.8±5.9	182.1±6.0
28.3±0.9	13.6±1.2	71.5±2.8	38.1±1.8	125.2±3.2	78.6±2.4	174.3±3.5	125.6±1.9
83.7±2.8	72.5±4.3	158.1±5.2	114.6±7.4	223.1±11.2	168.2±5.8	348.8±10.7	299.9±9.4
71.2±2.7	59.8±1.3	164.2±11.0	128.8±2.7	240.7±10.9	181.9±5.2	294.8±15.4	253.6±8.2
29.8±1.2	15.4±1.0	65.9±1.4	36.9±1.4	127.1±2.3	79.2±2.3	177.7±2.4	120.2±3.3

to Reitan, such forces may reduce the potential for patient discomfort, tissue hyalinization, undermining resorption.

Initiation of orthodontic treatment with "leveling" archwires requires wires with great range to accommodate the usual malalignment of bracket slots in the untreated malocclusion. Low stiffness is advantageous so that the forces can be kept as gentle as possible.[5,13,14]

Based on the elastic property ratios of strength, stiffness, and range, two principal types of wires are suggested, i.e., either a multistranded stainless steel wire or a nitinol-type wire. The former capitalizes on conventional variable cross-section orthodontics, which was tabulated many years ago in an orthodontic textbook.[15] The latter makes use of variable modulus orthodontics.[1,5]

Previous studies have shown that multistranded stainless steel wires provide viable alternative to expensive titanium alloy wires. However, some studies have shown that titanium alloys are superior wires during aligning and leveling.[16,17]

In pursuit of a suitable initial archwire for leveling and aligning with proper load deflection behavior and to comprehend the abovementioned, the present study of activation and deactivation forces of initial archwires of 0.014-inch and 0.016-inch nickel–titanium, copper–nickel–titanium, and co-axial wires was taken up.

The parameters chosen for testing were activation and deactivation forces, which indicate the forces acting during orthodontic tooth movement.A large sample size was chosen for the *in-vitro* study which consisted of 96 archwires in order to minimize the error. All the wires were tested in the as-received conditions. The methodology adopted for this study was a modified three-point bending test, which was different from the previous studies.

The advantage in this methodology was that the jig was specially designed simulating the arch form, and the mid-span distance was selected based on the standard interincisor bracket distance. The archwire was secured at three points on the either side of the midspan simulating the incisor, canine, and first molar brackets. To simulate the oral conditions, Type III copper–nickel–titanium was selected for testing as these wires have TTR of 37°C, which is closer to oral temperature.

The results of the present study indicate that, in general, the deactivation forces are less compared to activation forces at any deflection for any wire, as there exists inevitable force dissipation. They also depict slightly higher standard deviations.

As the cross-section of the wire increases, i.e., from 0.014 to 0.016, the activation and deactivation forces also increase, irrespective of the wire material, as depicted in all the tables and graphs. This implies that the load-deflection forces are directly proportional to the cross-section of the wire.

Among the tested wires, 0.016 nickel–titanium showed the highest load-deflection forces, whereas 0.014 co-axial wires showed the least. These findings were similar to the previous studies.[16,17] Some studies have reported that the co-axial wires are viable alternative to the Titanium alloy,[18-20] but in this study, the activation and deactivation forces of co-axial wires were incomparable at any point of deflection.

In the context of the present study, less load-deflection forces and high percentage force loss at any deflection for co-axial wires could be attributed to the lack of shape memory, super elasticity, and may also be due to the deformation of the wire during deactivation. This also suggests that, for higher or lower deflections, co-axial wires may not be suitable as initial archwires because they undergo deformation at higher deflections and produces less than optimal orthodontic force for lower deflections such as at 0.5 mm. Copper–nickel–titanium as claimed by manufacturers should express relatively constant force at increased deflections. However, in the present study,

the activation and deactivation forces increased at higher deflections.

The force values of nickel–titanium, copper–nickel–titanium and co-axial wires were higher in this study compared to the previous studies.[21,22]

The abovementioned observations could be attributed to various factors such as variation in jig design, friction between the archwires, the securing screws and manual errors during the monitoring, and maintaining the temperature of artificial saliva during the testing procedures.The selection of initial archwire certainly depends on the type and severity of the malocclusions.[23,24]

In most of the clinical situations, 0.016 nickel–titanium can be used for initial leveling and aligning effectively due to its desired properties and load-deflection forces which was similar to those shown in earlier studies.[1] However, in few clinical situations such as in severe malocclusions and cases where large deflections are needed, initial wires should be selected wisely.

The sample size of the study was high at arch wires; however, the study if planned in an *in-vivo* set after some modifications can be more validated.

The results obtained in the study have shown that there is a variation, especially in relation to the co-axial wires, which clearly indicates that more amount of research is indicated. No systematic review exists and a meta-analysis into the behavior pattern of wires is required, which will indicate a suitable archwire for the needed phase.

CONCLUSION

From the study, it can be concluded that activation (loading) and deactivation (unloading) forces are higher for nickel–titanium followed by copper–nickel–titanium and co-axial wires. Hence, nickel–titanium archwires are wires of choice for severely malpositioned and crowded malocclusions. It can also be concluded that stiffness is directly proportional to the size of the wire. In this study, 0.016 nickel–titanium wires had the higher stiffness compared to other wires of the same size and 0.014 sized nickel–titanium and other archwires.

The study also showed that the amount of percentage force loss was more for co-axial wire, indicating that these wires are not ideal wires for initial leveling and aligning. The high amount of force loss during activation and deactivation or co-axial wires are due to lack of super elastic properties and shape memory compared to nickel–titanium and copper–nickel–titanium wires.

An important conclusion form the study was that, in cases of severely malpositioned teeth, 0.016 nickel–titanium wire could be the wire of choice due to its unique properties of super elasticity and shape memory.

However, it has to be said that no ideal archwire exists and further research has to be conducted at different temperatures, and an *in-vivo* study pattern, which was a limitation here, needs to be conducted in order to understand the correct behavior of the archwire.

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Conflicts of interest

There are no conflicts of interest.

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