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MECHANICAL ANALYSIS OF ORTHODONTIC WIRES

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ABSTRACT: Orthodontic tooth movement is a physiologic response to externally applied forces; the motive forces are primarily mechanical. The optimal application of orthodontic force enables maximum movement of teeth with minimal irreversible damage of the periodontal ligament (PDL), alveolar bone, and teeth. Since arch wires are the main force system in orthodontics, it is important in clinical practice that they deliver appropriate, predictable and repeatable forces during treatment. These specialized wires even promise shape memory properties and the possibility of super elastic behavior, which significantly impacts clinical practices. Since, standard stainless steel and titanium arch wires are still the materials of choice in many stages of treatment. They provide an attractive combination of stiffness, resilience and formability.

However, clinical practitioners have commented on the variability of arch wire behavior for years. Inconsistent arch wire properties can contribute to unpredictable treatment duration and results.

This paper examines the mechanical and physical characteristics of stainless steel and titanium wires to quantify their variability in engineering terms. From the results for both types of wires, the testing method provides the information required by designers wishing to improve the arch wire properties and provide valuable information to clinicians for their practice.

Keywords: orthodontics, mechanical properties, tensile properties, orthodontic wires, titanium, stainless steel.

INTRODUCTION:

Teeth move through alveolar bone by strains generated through orthodontic appliances. The possible exception of tooth drift, orthodontic tooth movement is accompanied by minor reversible injury to the tooth-supporting tissues [1], superimposed on the physiologic adaptation of alveolar bone to mechanical strains. Relevant inflammatory mechanisms need to be considered along with skeletal mechanotransduction for a full understanding of orthodontic tooth movement. The clinical picture of orthodontic tooth movement consists of three phases: an initial and instantaneous tooth displacement; delay, where no visible movement occurs; and a period of linear tooth movement. The applied forces create strains in the tooth-supporting tissues that manifest immediately and can be categorized as compressive and tensile. The initiating inflammatory event at compression sites is caused by constriction of the periodontal ligament microvasculature, resulting in a focal necrosis, known by its histological appearance as hyalinization, and compensatory hyperemia in the adjacent periodontal ligament and pulpal vessels. These necrotic sites release various chemo-attractants that draw giant, phagocytic, multi-nucleated, tartrate-resistant acid-phosphates-positive cells to the periphery of the necrotic periodontal ligament. The cells resorb the necrotic periodontal ligament, as well as the underlying bone and cementum. Osteoclasts are recruited from the adjacent marrow spaces [2]. Until these cells can be recruited and the necrosis removed, tooth movement is impeded, resulting in the clinical manifestation of a delay period. This is followed by deposition of new cementum [3, 4], pulpal secondary dentin, and bone [5] in the vicinity of resorption sites.

The inflammatory responses to tension may be strain-dependent, since tensile strains of low magnitude are anti-inflammatory and induce magnitude-dependent anabolic signals in osteoblast-like periodontal ligament cells, culminating in the regulation of inflammatory gene transcription [6]. High tensile strains act as pro-inflammatory stimuli and increase the expression of inflammatory cytokines [7]. One issue that, at first, seems paradoxical is the observation that compression sites in orthodontic tooth movement are primarily resorptive, while the tensile sites are osteogenic. There are two possible explanations for these differences. First, compression sites have a tissue injury component superimposed on physiological transduction, with the former producing inflammatory products that are primarily resorptive and stimulate cells to remove the injured tissue. Second, resorption at compression sites in tooth movement perceived as lowering of the normal strain from the functioning periodontal ligament, while osteogenesis at tension sites is a reflection of loading of the principal fibers of the periodontal ligament, it is accompanied by strains in the alveolar process transmitted either through the principal fibers of the periodontal ligament or by direct impingement of the tooth root on the alveolar bone [8].

There are two fundamental requirements for orthodontic tooth movement to occur:

(1) Soft tissue is required, intervening between tooth structure and alveolar bone, which plays an important role in regulating the remodeling of adjacent tissues; and

(2) A bone turnover that is temporally and spatially regulated to facilitate specific translocations of teeth through alveolar bone [8].

There are four essential interrelated steps in the transduction of mechanical signals by tissues: sensing the mechanical signal by the cells, transduction of this mechanical signal into one that is biochemical, then transmission of the biochemical signal to the effector cells, and finally the effector cell response [8].

The problem of the transmission of forces and stresses on teeth and surrounding tissue is complex due to the inhomogeneous character of the structures of which they are constructed and irregularity of contour, and also their external form and complex internal morphology. Each tooth consists of several different tissues: enamel, dentine, pulp, cement and the periodontal ligament connected to surrounding bone as shown in figure (1). Each of the above tissues has essentially different characteristics and properties. A true and firm mechanical body, such as the tooth, changes its form under the influence of external forces, during which additional internal forces take place among the molecules in the interior of the body. The changed form is defined as deformation, while the additional forces among the molecules are defined as stress.

Final movement of the body, as a whole, or its particular parts, occurs as a result of the external forces. All these factors (forces, deformation, stress and movements) are mutually connected [8].

Orthodontic treatment works by applying steady pressure to the teeth, moving them gently and gradually into new positions, figure (2). This is possible because bone even though if feels hard, like ivory is actually flexible. When a tooth is pushed in a certain direction, the bone in front of it gives way under the pressure, and new bones form behind it to fill the gap. Each wire or rubber band in braces works like a spring –if stretched out and let go, it returns to its original position. The dentist takes advantage of these forces to move teeth .The wires or rubber bands are stretched, and as they return to their original positions ,they apply steady pressure on teeth , causing them to move [9].

TYPES OF ORTHODONTIC WIRES:

The arch wires are the main force system in orthodontics, they have been experimentally tested many times over. In some cases, the objective of the study was to measure the mechanical properties of wires, thus quantifying their mechanical behavior. This characterization eliminates some of the subjectivity involved in choosing the appropriate wires at various stages of treatment. Many other studies have attempted to compare and contrast the properties of different alloys, thus determining the advantages and disadvantages of newer materials.

There are some requirements that must exist in the wires used in orthodontics and these are:

- Non-corrosive.
- Easily formed.
- Maintain shape.
- Controlled and reproducible force delivery.

All orthodontic wires are made from an alloy and they are capable of bending a large distance without becoming permanently deformed [10].

MATERIAL AND METHODS:

The square stainless steel and titanium orthodontic arch wires of sizes 0.4 x0.4 mm for the stainless steel and 0.4 x 0.4 mm for the titanium were tested to find the modulus of elasticity (E), yield strength (S_y), ultimate tensile strength (UTS), and modulus of resilience (Ur).

Tensile properties

Elastic modulus is one of the critical factors in determining the clinical performance of orthodontic archwires.

A standard tensile test was performed on stainless steel and titanium wires in a WP 300 universal Material Tester as shown in Fig. (3). The gauge length of each specimen was set to 100 mm. The crosshead speed was set to 1mm/min and the axial load and axial strain data obtained were plotted as stress-strain curves.

The slope of the linear portion of the curves yielded E, as given by the following formula [11]:

$$E = \frac{\sigma}{\varepsilon}$$

Where σ is the induced stress and ε is the axial strain. S_y was measured at a 0.2% strain offset; and, UTS was found from the maximum axial stress.
The modulus of resilience, Ur, is the area under the stress-strain curve up to yielding. It represents the amount of recoverable energy during unloading. This energy is translated into applied forces on the bracket, which is the result of the wire returning to its formed state. It is calculated from [11]:

$$Ur = \frac{1}{2}\sigma y \ \varepsilon y$$

, where σy is the yield stress and εy the yield strain.

The objective of the test is to measure the mechanical properties of both types of orthodontic wires: modulus of elasticity (E), yield strength (S_y), ultimate tensile strength (UTS), modulus of resilience (U_r). In this study I shed light to explain how the modulus of elasticity for both types of wires that are used in orthodontics plays an important role in determining the clinical performance of orthodontic arch wires.

RESULTS:

Tensile Properties

The load deflection curves obtained by the tensile test were plotted as stress-strain curves for each specimen as shown in figures 4 & 5. The mechanical properties obtained from these curves are summarized in Table 1.
For a comparison with other researchers results the following results were found: tensional elastic modulus of 193 GPa and ultimate strength of 1300 MPa. Krishnan and Kumar [12] have also tested stainless steel 0.017 x 0.025 in wires in tension to obtain the elastic modulus and ultimate tensile strength. The following results were found: elastic modulus of 170 ± 20 GPa, and ultimate strength of 2100 ± 40 MPa. Verstrynge, Humbeeck and Willems [13] have also tested stainless steel 0.017 x 0.025 in wires in tension to obtain the tensional elastic modulus and ultimate tensile strength. The following results were found: tensional elastic modulus of 166 ± 1 GPa and ultimate strength of 1986 MPa. Rucker and Kusy [14] have tested stainless steel wires, diameter of 0.012 in, in tension, to find the ultimate tensile strength and yield strength, and in bending, to find the elastic modulus. The following results were found: elastic modulus of 198 GPa and ultimate strength of 2280 ± 80 MPa. Vena et.al.[15] have also tested stainless steel 0.017 x 0.025 in wires in tension to obtain the elastic modulus and ultimate tensile strength. The following results were found: elastic modulus of 168.4 GPa, and ultimate strength of 2098.1 MPa. Also Vena et.al.[15] have tested stainless steel but with a cross section 0.019 x 0.025 in wires in tension to obtain the elastic modulus and ultimate tensile strength. The following results were found: elastic modulus of 169.8 GPa, and ultimate strength of 2041.7 MPa.

The mechanical properties taken from this study could serve as a basis for comparison between different types of wires. It can be said that fracture does not occur at the same stress-strain conditions for samples. However, it is experimentally difficult to predict the location of fracture of a specimen that does not have a “dog bone” shape. Since the main weakness of the specimen occurs in the grips, due to stress-concentration, fracture will become very sensitive to testing conditions.

The main governing body in orthodontics, the American Dental Association (ADA) [16] lists a number of testing methods to ensure that all variables remain constant from one study to another. Although this testing method may provide practitioners with clinically relevant information on the mechanical behavior of arch wires, it does not provide information on their mechanical properties. These properties could serve as a basis for comparison between different types of wires, which served as the purpose for this study.

DISCUSSION:

The main objective of the study was to examine the variability of stainless steel arch wires with respect to mechanical properties and cross-section dimensions. Two different material type, stainless steel and titanium specimen, were chosen to be studied.

Stainless Steel 0.4x0.4 mm:

A tensile test is the most common mechanical stress-strain test, where a specimen is axially loaded in tension, under a constant strain rate, until fracture. The resulting stress-strain curve yields the elastic and plastic behavior of the material. The strain hardening that occurs during the manufacturing process gives the wires an increase in strength and, by increasing the yield strength, increases the elastic region of the stress-strain curve. As a result, the wire requires higher stresses to produce a plastic deformation, indicating that a larger amount of recoverable energy can be stored during loading. This is evident by the large value of U_r, which shows the high spring back of the wires. However, the resultant material becomes more brittle, and as a consequence cannot withstand as much plastic deformation before fracture. Therefore, the manufacturing process increases the spring back of the wires but decreases their formability.

The surface quality of arch wires is very important, as it determines corrosion resistance, biocompatibility and frictions characteristics, the latter having large effects on the force transmission between arch wire and bracket [15].

These irregularities will act as stress raisers, which will weaken the material and make it more susceptible to fracture. In clinical practice, wires are constantly tightened and adjusted during treatment, which could result in plastic deformations. Consequently, the surface roughness would increase and would result in an increase in frictional losses.

Another effective way to characterize a material is to identify its fracture mode. Although the wires did not show a very large percent elongation during the tensile test, the fracture surface is at least typical of a moderately ductile material. This shows that the material does not fracture immediately after reaching its yield stress, but does show a slight amount of plastic deformation. In clinical practice, this means that the wires will display plastic deformation before they fail.

Titanium 0.4x0.4 mm:

The tensile stress-strain result of titanium wire shows a higher elastic material with a large amount of springback, than stainless steel, with a slightly little brittle nature.

Table-2 shows that there is variability in the tensile properties for stainless steel wires. Some differences arises during testing.

The results from the mechanical testing indicate that the wires have very consistent mechanical properties along their lengths. This confirms that, for clinical applications, the wires of titanium should be used at the beginning of treatment, when severe misalignments require large activation, and the wires of stainless steel should be used in the later stages of treatment, where applied forces are less significant.

CONCLUSIONS:

Optimal use of orthodontic wires can be made by carefully selecting the appropriate wire type and size to meet the demands of a particular clinical situation.

The modulus of elasticity of the orthodontic wires was different and can result in significant computational errors in orthodontic appliance mechanics. Results showed that, in general, diameters, elastic moduli, yield strength, tensile strength are important parameters for evaluating the mechanical properties and physical characteristics between materials; the wire diameter has some influence on the parameters.

This work shows that variability does exist in reference to some mechanical properties between this study and other values reported in other literature. Variability of the mechanical properties of the same material could be caused by inconsistent manufacturing methods; this should be taken in consideration for the clinicians that variability can occur and influence orthodontic results. Furthermore, from the results for both types of wires, the testing methods used show high reliability and the parameters selected for evaluation provide the information that could be necessary in the evaluation and comparison of different wire types in future studies.

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Table-1: Tension Test Results; UTS and E for stainless steel and titanium material.

Material Type	**UTS** (MPa)	**E** (GPa)
Stainless steel	1300	193
titanium	1615	179

Table-2: A comparison Tension Test Results; for different studies.

Stainless steel	**The cross sectional size**	**UTS** (MPa)	**E** (GPa)
The current study	0.4x0.4 mm	1300	193
Krishnan and Kumar [12]	0.017x0.025 in	2100±40	170±20
Verstrynge, Humbeeck and Willems [13]	0.017x0.025 in	1986	166±1
Rucker and Kusy [14]	diameter of 0.012 in	2280±80	198
Vena et.al.[15]	0.017x0.025 in	2098.1	168.4
Vena et.al.[15]	0.019x0.025 in	2041.7	169.8

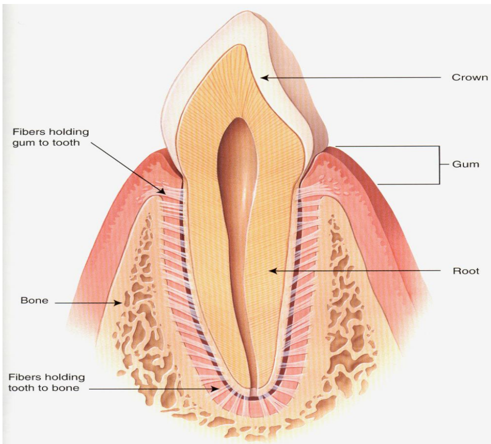

Fig.(1) : the tissues in the tooth [9].

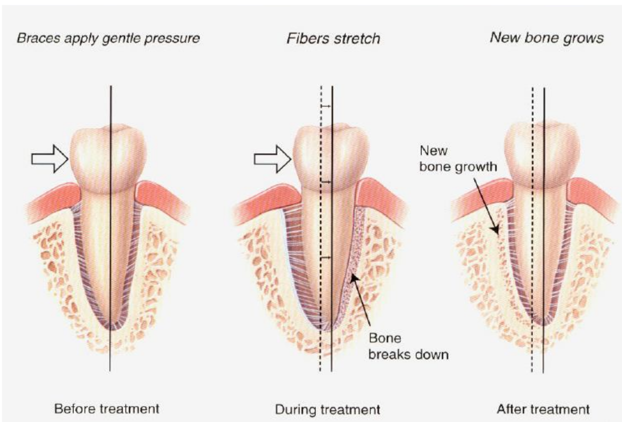

Fig.(2): The mechanism of tooth movement [9].

Fig.(3): The universal WP 300 Tensile testing machine**.**

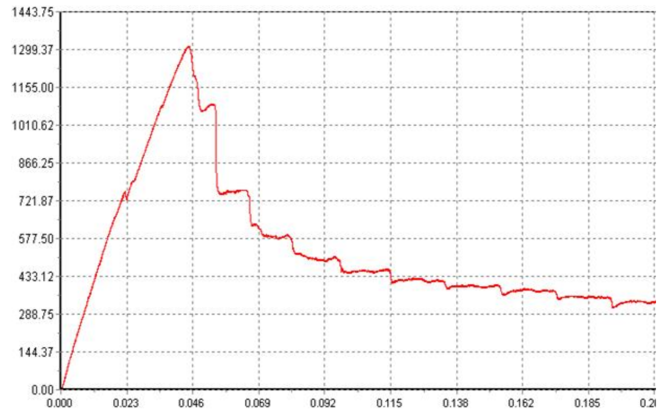

Fig.(4): Stress-Strain curve of stainless steel specimen the x-axis the strain in the specimen and the y-axis stress (MP/mm^2).

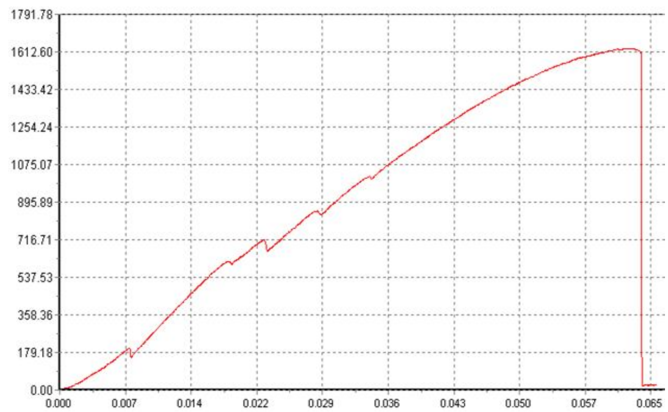

Fig.(5): Stress-Strain curve of titanium specimen the x-axis the strain in the specimen and the y-axis stress (MP/mm^2).

التحليل الميكانيكي لأسلاك تقويم الأسنان

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الخلاصة:

حركة الأسنان أثناء التقويم تمثل عملية استجابة فسلجية لقوى خارجية، والتي غالبا ما تكون هذه القوى ميكانيكية. التطبيق الأمثل لقوى التقويم تسمح بحركة قصوى للأسنان مصاحب بضرر بسيط للأسنان،الأربطة المحيطة بالأسنان، وعظم السنخى.

تعتبر الأسلاك المتقوسة المصدر الأساسي لتسليط القوة في جهاز التقويم. فمن خلاله يمكن أن نسلط قوة مناسبة خلال جلسات العلاج حيث ان لشكل السلك وسلوكه المرن تأثير واضح على فترة العلاج. إلى حد هذه الفترة تعتبر أسلاك التيتانيوم والستينلس ستيل الخيار الأمثل في المراحل العلاجية المختلفة لتقويم الأسنان وذلك لأنها تجمع بين خاصية الصلابة، المرونة، وقابلية التشكيل في جهاز التقويم.

من ناحية أخرى، يعاني أطباء الأسنان وخلال سنوات من عدم القدرة على تحديد فترة التقويم حيث يعزى ذلك لتضارب الخواص الميكانيكية للأسلاك المستخدمة في التقويم من سلك لآخر.

انطلاقا من الدور الذي يؤديه السلك في أداء جهاز التقويم انشأ هذا البحث والذي اهتم بدراسة الخواص الميكانيكية لأسلاك التيتانيوم والستينلس ستيل لتحديد المتغيرات التي تحدث ضمن الحدود الهندسية. كما حددت النتائج الخواص الميكانيكية الواجب توفرها في ألأسلاك المستخدمة في التقويم.