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Relative Stiffness of Beta Titanium Archwires

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ABSTRACT

The number of vendors of beta titanium (TMA™ type) wires has recently expanded dramatically. Samples of all the available sizes were obtained from most of the major beta titanium vendors. Three new square sizes are now also available. A total of 34 wire samples were tested for stiffness using the new ADA three-point wire-testing jig. Results show that not all beta titanium wires have the same stiffness. The range of variation was from small to large depending on the nominal wire size. There was also a spread of 1.67% to 4.27% in the standard deviation of the average stiffness from vendor to vendor. Burstone's Vari-modulus of elasticity is discussed along with how the properties of beta titanium could be integrated with his concepts. Strategies for using the new sizes of rectangular beta titanium wires for torque are discussed, including their use as working wires and finishing wires. The possibility of replacing stainless steel wires in the final finishing stages is also discussed.

KEY WORDS: Modulus of elasticity, Stiffness, Nominal torque, Slot play, Target torque, Custom prescription torque, Hypertorque.

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INTRODUCTION Return to TOC

Teeth respond to a consistent force on their periodontal membrane by moving. In orthodontics, the source of the force is usually a deflected spring or archwire. The amount of force generated is dictated by the spring rate and by the distance of the clinical activation. The spring rate of any spring, including archwires, is determined by a fixed constant, the moment of inertia, determined by the spring's cross section, and the modulus of elasticity of the spring's alloy.

How do we manipulate the stiffness of wires? The modulus of elasticity is a commonly used engineering term that describes a material's resistance to bending, ie, stiffness. It is the major factor determining the ultimate stiffness of an orthodontic wire. The original edgewise appliance used gold alloy wires with a modulus of elasticity of approximately 15×10^6 pounds per square inch (P.S.I.).[1] This low modulus allowed full archwire engagement in the 0.022-inch slot without undue forces. These gold wires, however, were very expensive and had such a low yield point that they were often inadvertently deformed in the mouth by occlusal activity. The yield point is the maximum amount of stress that a wire can endure without becoming permanently bent (Figure 1). A low–yield point wire is easily bent, whereas a high–yield point wire requires more force to create a permanent bend. Yield point and strength are very similar terms.

In the 1940s, austenitic stainless steel was developed. Stainless steel's modulus of elasticity was approximately 23×10^6 P.S.I., which is roughly 50% stiffer than gold (Table 1). Stainless steel also has a higher yield point than gold. A higher modulus of elasticity required that orthodontic wires be downsized to avoid creating excessive clinical forces. Stainless steel, however, was less expensive, formed well, and could be joined either by spot welding or soldering. Despite its increased stiffness, it became the orthodontic clinician's wire of choice over gold.

Elgiloy™, a chromium cobalt alloy, was originally developed for watch springs. It was adapted for orthodontic use later. Its modulus of elasticity is approximately 28 × 10^6 P.S.I., which is slightly stiffer than stainless steel.[1] The advantage of Elgiloy™ was that it could be manufactured with various tempers. A low-temper wire has a low yield point making it easy to bend complicated loops. After the wire configuration was formed, the wire could be heat treated, using precipitation hardening, which increased the yield point to a level similar to stainless steel. Heat treating only affects the yield point. Heat treating does not change the stiffness. Also, Elgiloy™ can be soldered. The clinical bending characteristics of Elgiloy™, stiffness and formability, are quite similar to those of stainless steel.

In 1978, Andreasen[2] introduced a stoichiometric nickel-titanium alloy called Nitinol™ to the orthodontic world. The modulus of elasticity of the original, work-hardened martensitic version was 4.8 × 10^6 P.S.I. The lower modulus of elasticity (21% of stainless steel) enabled very large clinical deflections before excessive forces were created. Nitinol™, however, is not formable and tends to break easily when bends are attempted. Its fracture point is quite close to its yield point (Figure 1—fracture point #1), and it can neither be soldered nor spot welded.

Pure titanium is normally found in a hexagonal close-packed (HCP) lattice form. The HCP lattice titanium demonstrates only one-third the yield strength of stainless steel. If it is heated above 1625°F, the structure changes to a body-centered cubic lattice referred to as the " beta phase." Upon cooling, the lattice structure reverts back to the original, low–yield point, HCP alignment. The HCP yield point is even less than that of gold wire, making it of no clinical use.

In the 1960s, an entirely different high-temperature form of titanium alloy became available.[3] Researchers found that if titanium was heated beyond 1625°F in the presence of modifying elements such as chromium, cobalt, columbium (niobium), copper, iron, manganese, molybdenum, nickel, tantalum, or vanadium and then cooled, the beta lattice structure was maintained after cooling. These metastable alloys are referred to as beta-stabilized titanium. These beta-phase alloys have a modulus of elasticity varying from 8 to 16 × 10^6 P.S.I. depending on the specifics of alloying, heat treating, and work hardening done during the manufacturing process. The modulus of elasticity of these alloys can be varied!

In 1979, Goldberg and Burstone[3] presented their preliminary work investigating the feasibility of using beta-stabilized titanium alloys for orthodontic applications. After much testing, Goldberg and Burstone[3] determined that, with proper processing, 11% molybdenum, 6% zirconium, and 4% tin beta titanium alloy would be usable in orthodontics. The modulus of elasticity of their wire, later called TMA[‡]™ (Titanium Molybdenum Alloy) was 9.4 × 10^6 P.S.I. or 41% that of stainless steel. It could be bent 105% further or twice the distance of stainless steel before reaching its yield point. Beta titanium cannot be soldered, but it can be welded to itself with minimal strength change.[1,5]

The stiffness of any orthodontic wire is determined by the geometry of the wire's cross section, its moment of inertia, and the alloy's modulus of elasticity. In the past, clinical wire stiffness was modified only by changing the cross-sectional geometry of the wire or by increasing the interbracket wire length by incorporating complex loops. The modulus of elasticity remained constant because stainless steel was the only alloy being used. Larger stainless steel wires would be stiffer, whereas smaller stainless steel wires would be less stiff. If the smaller wires were still too stiff, interbracket loops were incorporated or multiple strand wires were used.

Although Burstone mentioned the concept of changing the modulus in 1979,[3] he waited until 1981, when TMA[‡]™ became available, to formally introduce the concept of "Vari-modulus orthodontics."[4] Burstone[4] proposed that changing the wire alloy, not just changing the wire size, could also modify wire stiffness. Because each alloy has a different modulus of elasticity, each alloy would produce a wire of a different stiffness. This dramatically expanded our options for controlling stiffness. Using a lower modulus alloy would enable the use of a larger wire earlier in the succession of leveling wires without increasing stiffness. Ultimately, a succession of stiffer and stiffer full-fit wires could then be used. High-modulus stainless steel and chromium cobalt alloys precluded the use of full-fit wires especially in the earlier leveling stages. Now, stiffness could be modified by changing the stiffness of the alloy, changing the cross-sectional geometry of the wire, or by a combination of the two.

Burstone's variable-modulus concept started with a very flexible full-fit–leveling wire (very low modulus with high deflections). The next wire would not need to be deflected as much, requiring only a higher modulus while still keeping the wire full fitting. This would progress until finishing with a stainless steel wire.

The stiffness of any wire can be measured and compared. Johnson and Lee[7] published stiffness tables, organized both by ascending stiffness and by ascending wire size, in 1989. These tables enable the clinician to first decide on the clinical stiffness required and then select the combination of wire size and alloy that would be the most advantageous for the specific stiffness range.

Burstone's variable-modulus idea was before its time in 1981 because there were not very many well-developed alloy families to choose from. Stainless steel was still the primary alloy for orthodontics with some softer twisted and braided versions of stainless steel available.

Gold and chromium cobalt had been essentially passed over. Only nickel titanium (NiTi) alloys made of work-hardened martinsite were available. Because of the limited variety of wire stiffnesses available, smooth jumps in stiffness could not be made from leveling to working to finishing wires.

Over the years, the variable-modulus concept has slowly caught on as more and more varieties of an alloy family with multiple moduli of elasticity have been brought to the orthodontic marketplace. Today, we have expanded families of stainless steel, chromium cobalt, three generations of nickel titanium alloys and, more recently, beta titanium. Our choices have been increased even more because of the expanded numbers of twisted, woven, and coaxial versions of NiTi and stainless steel alloys now available. However, we always need current data about the clinical stiffness of all available wires so that we may pick successive archwires intelligently.

Archwire succession

Orthodontic wires need to produce a force that will move teeth efficiently. If the force is too high, movement will be slow, nonexistent, or painful. If it is too low, movement will be slow or nonexistent. Fortunately, there is a range of pressures that creates efficient tooth movement. I refer to this as the speed zone. Every wire, at the proper amount of activation, will produce forces that will pass through the speed zone during its deactivation. As the tooth moves, archwire deflection diminishes decreasing the force delivered. This relationship continues until the ever-lessening force is below the speed zone and no longer effective. Thus, because of subsequent tooth movement, no wire can continuously deliver a force that stays in the speed zone.

Because no one wire will continuously deliver the optimum force, a strategy of changing wires has been developed. To maintain efficient movement in the speed zone, a succession of stiffer and stiffer wires is needed as teeth move. The successive wires are selected based on their stiffness. The first wire is selected so that when it is tied into the most severely displaced bracket, it will deliver a force that is at the highest acceptable level of the speed zone. As the tooth moves, the force level delivered will decline and approach the lowest level of the speed zone. The initial wire needs to be replaced as its force level becomes suboptimal. The replacement wire should deliver, at the same deflection as the departing wire, a greater force that is at the highest level of the speed zone. As the second wire's force level drops to a suboptimum level, it should also be replaced in the same fashion. Based on our clinical experience, a replacement wire's stiffness usually needs to be two to three times higher than the stiffness of the wire being replaced. For this strategy to work, a large inventory of available wire stiffnesses is needed for efficient tooth movement.

When Burstone introduced his version of beta titanium (Titanium Molybdenum Alloy[*]) to the orthodontic profession in 1981, he provided the missing link in the modulus of elasticity progression outlined in Variable-Modulus Orthodontics.[4] Its modulus of elasticity was almost twice that of NiTi™ and less than half that of stainless steel. It was the natural step between a soft, unadjustable NiTi™ wire and a very stiff, formable stainless steel wire.

Burstone patented the use of beta titanium wire in orthodontics, and TMA™ was produced and sold by Ormco under a royalty agreeme with Burstone. Because of the patent and limited royalty agreements, only two versions of the beta titanium family have been available in the past (TMA[‡]™ and Titanium Niobium[‡]™). Recently, the patent for the orthodontic use of beta titanium expired, thus allowing this alloy family to expand drastically. The stiffness of these newly available wires is unknown.

The original TMA[‡]™ only came in four sizes, 0.016 inch × 0.022 inch, 0.017 inch × 0.025 inch, 0.0175 inch × 0.0175 inch, and 0.021 inch × 0.025 inch. Beta titanium is now available in seven different sizes from eight different vendors. The new square 0.018 inch × 0.018 inch, 0.020 inch × 0.020 inch, and 0.021 inch × 0.021 inch sizes present some new and interesting treatment possibilities.

The purpose of this study is to

- test the stiffness of all available wires in the newly expanded beta titanium family;
- compare and rank their stiffness to facilitate clinical wire selection;
- illustrate some clinical applications of this alloy.

METHODS AND MATERIALS Return to TOC

All the major wire vendors advertising beta titanium wires were contacted, and eight different vendors supplied 34 different test-wire lots (Table 2). There are six different nominal sizes of beta titanium wires represented. Three samples of Ormco's Titanium Niobium™ were included because it is a beta titanium alloy with bending characteristics similar to those of TMA™ except for a lower yield point (F. Farzin-Nia, personal conversation).

Most of the sample lots received contained 10 preformed arch blanks. The shape of the arch form blanks was randomly selected from the different arch forms available. All the wire samples provided were commercially available for shipment to any customer. Each lot was repackaged and labeled only with its nominal size and its randomly assigned test code number. All wire testing and data processing was done on a blind basis.

Five arch blanks were randomly selected from each 10-blank lot. A 20-mm test section was cut from the most distal portion of each of the five selected arch blanks. These five test segments were stored at 37°C for at least 10 minutes before testing to ensure that the wires had stabilized their temperature at 37°C.

All wires were tested using the new ADA-standardized three-point wire-testing jig.[8] The unsupported wire span was 12 mm long. The calibrated test plunger was centered six mm from the wire supports (Figure 2). The test plunger's load cell data and movement data were connected to an *x-y* graphic plotter. As the plunger traveled, deflecting the test specimen, a plot was produced by the *x-y* plotter reflecting both the bending resistance of the wire sample and the extent of plunger travel. Force was measured in grams, whereas the resulting deflection was measured in hundredths of millimeters. The entire test jig was contained inside a clear lucite enclosure with the temperature maintained at 37 ± 1°C, which replicated the normal intraoral temperature (Figure 3).

Each sample was placed in the testing jig and slowly deflected, producing the loading plot. Loading was continued until the plunger force reached 300 g. Loading was stopped at or just past 300 g. The plunger travel was then reversed, gradually reducing the plunger force back to zero g, thus producing the unloading plots.

The loading curve was ignored. A straight line representing the average unloading curve was drawn directly on the test plot (Figure 4). The individual slope for each sample was calculated in grams per millimeter of deflection. The five slopes were averaged to produce the mean. The standard deviation of the mean was then calculated. Also, because the standard deviations varied in magnitude as the means varied, each standard deviation was divided by its mean, producing a second standard deviation stated as the percentage of the actual wire stiffness. All the results are shown in Tables 2 and 3.

RESULTS Return to TOC

All test plots demonstrated almost linear loading and unloading curves. No permanent deformation or set was observed. No superelasticity was demonstrated (Figure 4).

0.016 inch × 0.022 inch

The wire stiffness values ranged over 162 g/mm, going from 524 g/mm (G&H Wire Co., Greenwood, Ind.) to 687 g/mm (Highland Metals, San Jose, Calif.). The average stiffness was 603 g/mm. Ormco's (ORMCO Corp., Glendora, Calif.) TMA stiffness was 8.3% below the group average at 553 g/mm. Unitek (Unitek Corp. Monrovia, Calif.) had the lowest standard deviation percent, at 1.62%, whereas Lancer demonstrated the highest, at 5.93% (Table 2).

0.017 inch × 0.025 inch

The wire stiffness values ranged over 181 g/mm from 652 g/mm (Unitek) to 833 g/mm (GAC International, Islandia, NY). The average stiffness was 735 g/mm. Ormco's TMA™ tested at 781 g/mm, which was 9.4% stiffer than the group average. Lancer (Lancer Orthodontic San Marcos, Calif.) had the lowest standard deviation percent, at 0.98%, whereas Ormco's TMA™ was the highest.

0.0175 inch × 0.0175 inch

There were only two 0.0175 inch × 0.0175 inch samples available for testing. Unitek's wire, at 701 g/mm, was 22.6% stiffer than Ormco's TMA™, at 573 g/mm.

0.018 inch × 0.018 inch

The wire stiffness values ranged over 96 g/mm, going from 665 g/mm (G&H) to 761 g/mm (Highland Metals). The average stiffness was 712 g/mm. Highland Metals had the lowest standard deviation percent, at 1.06%, whereas GAC's was 4.97%. TMA™ is not manufacture in this wire size.

0.020 inch × 0.020 inch

Highland Metals is the sole vendor of this size. Their sample arrived too late to be tested.

0.021 inch × 0.021 inch

There were only two samples available for testing. GAC's wire, at 1020 g/mm, was only 1% stiffer than Unitek's wire, at 998 g/mm.

0.021 inch × 0.025 inch

The wire stiffness values ranged over 83 g/mm, from Ormco's, at 1166 g/mm, up to GAC's, at 1340 g/mm. The average stiffness was 1238 g/mm. Ormco's TMA™ was 5.8% below the group average at 1166 g/mm.

The average stiffness of Titanium Niobium™ was 8.6% higher than TMA™ at 1266 g/mm, although Titanium Niobium™ is advertise being 80% the stiffness of TMA™.[9] Its yield point was designed to be less than that of TMA™, making it a finishing wire that was easy to adjust (F. Farzin-Nia, personal conversation). Yield points were not tested in this study.

DISCUSSION Return to TOC

Wire succession

Beta titanium wires fill an important niche in the progression of wire stiffnesses of orthodontic wires. Beta titanium wires are stiffer than unformable nickel titanium wires. They, however, are the softest formable alloy family and are significantly softer than stainless steel and chromium cobalt wires. The combination of high formability and low stiffness is considerably less than stainless steel making these wires the next logical step in wire progression after initial leveling with NiTi™ wires.

Initial leveling is usually done using preformed nickel titanium arch blanks that are relatively unadjustable. Their arch form is predetermined and fixed at the factory. If the appropriate arch form is not available, the clinician cannot change it and must accept a best-fit compromise.

In addition to arch form problems after initial leveling, bracket repositioning and releveling with preformed arch blanks often still leaves discrepancies that cannot be corrected by bracket repositioning. Further refinement and compensations are only possible by bending a formable wire, and NiTi™ wires are not very formable. The clinician must choose either a formable low stiffness/high-range beta titanium wire or a high stiffness/low-range stainless steel or chromium cobalt wire for a final detailing wire.

If you assume that succeeding archwires usually need to be two to three times the stiffness of the wire being replaced, the expanded range of stiffness of the second-generation beta titanium wires has given the clinician a broader range of stiffness choices for a given wire size. For example, in a 0.018-inch slot system when a full-slot–fit wire is desired, either the 0.0175 inch × 0.0175 inch or the 0.018 inch × 0.018 inch wires would do. The softest TMA™ available is the 0.0175 inch × 0.0175 inch wire, but other alternatives up to 35% stiffer (Highland Metals 0.018 inch × 0.018 inch) are available. We now have a broader stiffness range from which to choose!

This is particularly useful because my fourth wire, 0.0175 inch × 0.0175 inch TMA™, was only 44% stiffer than the 0.018 inch × 0.018 inch Lancer NiTi™ wire that I normally use as my third wire.[7] Now, the stiffer beta titaniums enable a wire progression with an increase in stiffness of up to 94% from my third to fourth wire (Table 2).

Working wires

Beta titanium archwires come from the manufacturer in many predetermined arch forms. Unlike NiTi™ wires, beta titanium archwires ca easily be modified to customize arch form and to make in-out, up-down, and torquing bends. Because beta titanium wires are much softer than stainless steel (42% the stiffness of steel) wires, they have over twice the range permitting larger wire adjustments before the speed zone is violated. Larger first- and second-order bends and torquing adjustments can be made, producing greater distances of tooth movement between adjustments.

This softness and increased range is especially useful when dealing with torque. The slot can be filled completely with a full-fit wire, reducing the slot play to a minimum while still delivering torque at a gentler, more constant rate (Table 4). The lower torque rate can also be used with traditional slot play amounts. If needed, torque can be custom adjusted by a greater angle at each appointment than is possible with a stainless steel wire with the same slot play.

Custom torque prescription (based on final finishing wire slot play)

Larry Andrews' straight wire concept[10,11] has now evolved from the original straight wire prescription past the preadjusted phase[12] into the customized preactivated phase. His straight wire torque prescription was designed to fit teeth that were already straight. Because of slot play, it has been only partly successful in correcting torque issues. A custom preactivated appliance is a preadjusted (straight wire) appliance with its prescription torque values selected to correct the specific torsional malposition of individual teeth. It does not use the same torque prescription for each and every patient. For example, if a root needs buccal torque, it will need one prescription (target torque minus slot play). If it needs lingual root torque, it will need quite a different one (target torque plus slot play).

Figure 4 graphically represents the amount of root torque produced in a straight wire appliance by a flat rectangular arch after it is twisted, seated in the deflected bracket slot, and released. Note the broad space on either side of the prescription torque where no torque is delivered. If the wire is twisted less than or equal to the slot play, the torqued wire does not lock, and no torsional force will be delivered to the bracket. The rectangular wire will act as if it were a round wire. Activation will produce a torquing force on a bracket only if the wire deflection exceeds the amount of slot play—in which case, as the root responds to the twisted wire, the final torquing action toward the prescription torque will stop at the slot play boundary, "target torque."

I define target torque as the prescription's designated stopping point of a torquing movement (Figure 5). It always differs from the nominal prescription torque by the amount of slot play encountered. Each prescription has two target torques. They are (1) the prescription torque plus the slot play and (2) the prescription torque minus the slot play.

The custom torque prescription for an individual tooth depends on the final crown torque position desired (target torque), the existing position of the crown (too lingual or too facial), and the amount of slot play the final finishing wire will experience while delivering torque. The total amount of clinical slot play is a function of the true size of the nominal bracket slot, the true size of the nominal wire, and the actual amount of corner rounding of the actual wire. Oversized slots, undersized wires, and large corner rounding increase slot play![12,13]

To illustrate these concepts, let us calculate the custom torque prescription needed for a division 2 upper central incisor positioned at −10° (excess labial root torque/insufficient buccal crown torque). The first question is, "What is the final target facial inclination desired when torquing is finished?" Andrew's norm of +7° is our target.[10,11†] The second question is, "Which way will the root need to be torqued from it present position, lingual or labial?" In this example, lingual root torque will be needed because the present crown inclination is −10° (anything +6° or less will need lingual root movement, whereas anything +8° or more will need labial root torque). The third question is " What is the slot play of the final finishing wire?"

Our example will use a 0.017 inch × 0.017 inch stainless steel wire with a total anticipated slot play of 9.5°. Because 9.5° of the twist in the 0.017 inch × 0.017 inch wire will be lost before it locks in the slot (because of slot play), the prescription needs to be increased 9.5° to compensate for the loss. Our custom nominal prescription torque that will produce a final facial target torque of +7° using a 0.017 inch × 0.017 inch archwire experiencing a 9.5°-slot play wire is +16.5° ([+7°] + [+9.5°]) (Figure 6). The slot play factor is added because the root is being torqued from the labial direction. If the root were to be torqued from the lingual direction, the slot play factor would have been subtracted, producing a nominal prescription torque of −2.5°. Although the nominal prescription torque is dependant on the direction of desired movement, the target torque (our treatment goal) is always the same!

In our division 2 example, when the flat "straight wire" archwire is tied in, it will need to be temporarily twisted to a total of 26.5° to gain entry into the +16.5° slot. Remember the facial of the crown was at −10°. As soon as the wire is released, it will lose 9.5° because of slot play, before locking. This will result in a residual torque activation of 17°. When the 17° activation has been spent, the root will have been torqued 17° lingually and will have stopped with the facial of the crown at +7°, our target torque (Figure 6)!

Hypertorque

Please note from the example above that as the root moves nearer to the target torque, the wire deflection has diminished, reducing the torsion and, probably, the rate of final root movement (Figure 7). If one wants to temporarily supercharge this final torquing action, one could increase the amount of effective activation by replacing the stainless steel finishing wire with a larger more flexible beta titanium wire with less slot play. Reducing the slot play temporarily moves the target torque past the desired final facial torque increasing the activation and speeding the final root movement (Figure 8). Hypertorque! I typically finish with a 0.017 inch × 0.017 inch stainless steel finishing arch (8.16° slot play).[14] Switching to a newly available 0.018 inch × 0.018 inch beta titanium (2.19° slot play) would temporarily add 6° of torque activation (Figure 8)! If the hypertorqued wire is left in too long, the root could be overtorqued past the prescription target torque. Up until that point, however, the softer, lower slot play wire would have delivered a more continuous and effective final torquing force to the desired target torque. By using a second generation full-fit beta titanium working wires and finishing with a higher slot play stainless steel wire, we can have a dual standard, efficient final torquing and finite control over the final crown position.

Final finishing wire

Stiffness comparisons indicate that beta titanium wires can often be used in lieu of stainless steel for final finishing wires. If the final first- and second-order detailing discrepancies are small, relatively stiff stainless steel finishing wires can be used. Small activations of a stiff wire will not build up excessive forces. Square stainless steel wires have lower loading rates than rectangular steel wires with comparable slot play. Steel ribbon arches have even lower loading rates.

If the required adjustments are larger, using the stiffer stainless steel wires often makes the resulting postadjustment forces excessive. Because beta titanium wires have ~ 40% the stiffness of stainless steel, they can be adjusted over twice the distance of stainless steel before building up equivalent forces. Beta titanium wires are excellent finishing wires where formability is required for large adjustments. The lower decay rate of adjustment forces will ensure more efficient root movement over a longer distance.

Beta titanium wire's lower stiffness enables the clinician to reduce slot play while applying torque without ending up with excessively high torsion forces. This is especially true for clinicians who use 0.022-inch slot appliances. A 0.019 inch × 0.025 inch wire is the most common torquing-finishing wire used with the 0.022-inch slot bracket. It has a slot play of 10.5° in a 0.022-inch slot. Its moment of inertia[‡] or torsional stiffness factor is 171,475. A 0.021 inch × 0.021 inch beta titanium wire has a slot play of 5° with a torsional stiffness factor of 81,682. It works with 5° less slot play, with half the torsional stiffness and with twice the available range of activation. Teeth can be torqued over a longer distance into their final position with more control (Table 5).

A 0.021 inch × 0.025 inch beta titanium wire has a slot play of 3.88° and a torsion stiffness factor of 97,240. It is slightly stiffer in torsion, but it still has a more accurate slot fit and is less stiff than the 0.019 inch × 0.022 inch stainless steel wire.

Beta titanium wires cannot be soldered. Hooks, springs, and other attachments made out of another piece of beta titanium can be spot welded with good success.[6] These comparisons suggest that beta titanium wires can often be an effective replacement for stainless steel final finishing wires.

CONCLUSIONS Return to TOC

- Beta Titanium wires are now readily available from multiple vendors under various brand names.
- Beta Titanium is now also available in 0.018 inch × 0.018 inch and 0.021 inch × 0.021 inch sizes.
- Beta titanium is available with a wider range of stiffness choices.
- Beta titanium has become a more useful class of wires for both working and finishing wires.
- In a 0.022 inch-slot appliance, beta titanium can replace stainless steel as a finishing wire.

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TABLES Return to TOC

TABLE 1. Modulus of Elasticity

NiTi™[a] (Original Nitinol)	4.8 × 10^6 pounds per square inch
TMA™[a]	9.4 × 10^6 pounds per square inch
Beta titanium[b]	8–16 × 10^6 pounds per square inch
Gold[a]	15 × 10^6 pounds per square inch
Stainless steel[a]	23 × 10^6 pounds per square inch
Elgiloy™[a] (chromium cobalt)	28 × 10^6 pounds per square inch

[a] From Burstone and Goldberg.[1]
[b] From Goldberg and Burstone.[3]

TABLE 2. Test Results by Vendor

Size (0.000 inch)	Mean Stiffness (g/mm)	Standard Deviation	% of Mean Standard Deviation
G & H Wire Company—Titanmoly™			
16 × 22	591	12.06	2.04
17 × 25	761	16.3	2.14
18 × 18	693	13.9	2.00
20 × 20	—[a]	—[a]	—[a]
21 × 25	1215	5.5	0.45
			Average 1.66
GAC—Resolve™			
16 × 22	645	32.2	4.99
17 × 25	673	7.39	1.10
18 × 18	730	36.3	4.97
21 × 21	1020	18.0	1.76
21 × 25	1340	10.7	0.86
			Average 2.74
Highland Metals—BetaTi™			
16 × 22	687	13.1	1.91
17 × 25	694	9.40	1.36
18 × 18	761	8.09	1.06
21 × 21	1190	30.6	2.57
			Average 1.73
Lancer Orthodontics—Full Range Titanium Alloy™			
16 × 22	640	37.5	5.88
17 × 25	834	8.18	0.98
25 × 25	1266	75.2	5.94
			Average 4.27
ORMCO Corp.—TMA™			
16 × 22	553	19.2	3.48
17 × 25	781	31.7	4.05
175 × 175	573	5.98	1.04
21 × 25	116	17.0	1.47
			Average 2.51
ORMCO Corp.—Titanium Niobium™			
16 × 22	617	36.6	5.93
17 × 25	744	15.5	2.08
21 × 25	1266	11.2	0.88
			Average 2.96
Ortho Organizers—Beta CNA™			
16 × 22	591	12.1	2.04
17 × 25	761	16.3	2.14
18 × 18	693	13.9	2.00
21 × 25	1215	5.5	0.45
			Average 1.67

			Average 1.67
Rocky Mountain—Bendaloy™			
16 × 22	591	15.2	2.57
17 × 25	733	28.7	3.92
			Average 3.25
Unitek—Beta III Titanium™			
16 × 22	580	9.40	1.62
17 × 25	652	6.80	1.04
175 × 175	701	11.0	1.57
21 × 21	998	17.0	1.70
21 × 25	1212	15.0	1.24
			Average 1.43

[a] Received too late to be tested.

TABLE 3. Test Results by Wire Size and Ascending Stiffness[a]

Size (0.000 inch)	Mean Stiffness (g/mm)	Standard Deviation	% of Mean Standard Deviation
0.016 × .022			
G&H	524	10.5	2.01
OTMA	553	19.2	3.48
OO	591	12.1	2.04
RM	591	15.2	2.57
UNI	580	9.40	1.62
OTN	617	36.6	5.93
LAN	637	37.5	5.88
GAC	645	32.2	4.99
HM	687	13.1	1.91
0.017 × 0.025			
UNI	652	6.80	1.04
GAC	673	7.39	1.10
HM	694	9.40	1.36
RM	7.33	28.7	3.92
G&H	741	14.5	1.95
OTN	744	15.5	2.08
OO	761	16.3	2.14
OTMA	781	31.7	4.05
LAN	833	8.18	0.98
0.021 × 0.025			
OTMA	1166	17.0	1.47
HM	1190	30.6	2.57
UNI	1212	15.0	1.24
OO	1215	5.5	0.45
G&H	1247	57.6	4.61
LAN	1266	75.2	5.94
OTN	1266	11.2	0.88
GAC	1340	10.7	0.86
0.0175 × 0.0175			
OTMA	573	5.98	1.04
UNI	701	11.0	1.57
0.018 × 0.018			
G&H	665	8.07	1.21
OO	693	13.9	2.00
GAC	730	36.3	4.97
HM	761	8.09	1.06
0.020 × 0.020			
G&H		(Submitted too late to be tested)	
0.021 × 0.021			
UNI	998	17.0	1.70

GAC	1020	18.0	1.76

[a] GAC indicates GAC; GH, G & H Wires; HM, Highland Metals; LAN, Lancer Orthodontics; OO, Ortho Organizers, San Marcos, Calif.; OTN, ORMCO—Titanium Niobium®; OTMA, ORMCO—TMA®; RM, Rocky Mountain, Denver, Colorado; UNI, Unitek.

TABLE 4. Relative Torsional Stiffness

Slot Play	Size (0.001 inch)	Alloy	Torsional Stiffness Factor
10.5	0.019 × 0.025	Stainless Steel	171,475
5.00	0.021 × 0.021	Resolve	81,682
3.88	0.021 × 0.025	TMA®	97,240
~1.0[a]	0.022 × 0.016	Stainless Steel	170,368
~1.0[a]	0.022 × 0.016	TMA®[b]	69,850

[a] Estimated slot play.
[b] No longer available.

TABLE 5. Archwire Slot Play[a]

Size (0.000 inch)	0.018 inch Slot	0.022 inch Slot
16 × 16	16.7	Spins
16 × 22	9.28	27.5
16 × 26	7.27	20.0
17 × 17	8.16	Spins
17 × 22	5.37	22.3
17 × 25	4.48	17.7
18 × 18	2.19	Spins
18 × 22	1.63	17.5
18 × 25	1.37	14.0
19 × 25		10.5
21 × 21		5.00
21 × 25		3.88
21.5 × 21.5		2.30
21.5 × 28		1.97
22 × 22		0.90

[a] Calculated assuming 0.003 inch rounded radius corners and exact (±0) slot height. Data from Meyer and Nelson.[16]

FIGURES Return to TOC

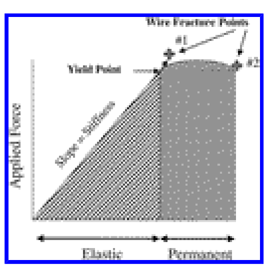

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FIGURE 1. Stress, strain, range, and formability. The yield point is where temporary elastic bending stops and permanent bending starts. The force at the yield point indicates the maximum clinical strength of the wire. The slope indicates the wire stiffness and is directly related to the modulus of elasticity. The width of the striped area indicates the elastic range. The width of the checkered area indicates the amount of formability. A brittle wire fractures quickly after the yield point (wire fracture point 1), whereas a formable wire can be permanently bent a long distance past the yield point before fracturing (wire fracture point 2)

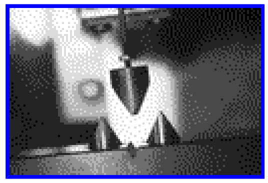

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FIGURE 2. Wire test jig. The center plunger is six mm from both supports. The total unsupported span is 12 mm. The ends of the wires beyond the jig are not supported

Click on thumbnail for full-sized image.

FIGURE 3. Photo of test housing. Test jig and plunger mechanism. Internal test temperature is maintained at 37°C ± 1°C

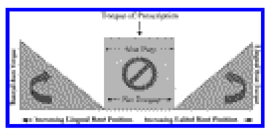

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FIGURE 4. Slot play–torque graph. This graph represents the amount of torquing force that is generated as a passive flat (zero torque) archwire is placed in a bracket. The right side of the graph represents the force (lingual root torque) generated as the root is found positioned more and more to the labial (with more and more lingual crown torque). The left half or the graph represents how much buccal or labial root torque is generated as the root is found progressively more and more to the lingual (with more and more labial crown torque). Note that there is no correctional torquing force developed until the tooth is displaced more than the extent of slot play. If the slot play has been exceeded (the rectangular arch is locked in the slot), the torque developed is a direct function of additional crown-root deflection in excess of the slot play. If the crown is deflected more than the extent of slot play, the resulting torsional correction toward the bracket prescription will stop short of the bracket prescription by the extent of the slot play. The slot play acts as a plus and minus buffer straddling the bracket slot's torque, making the actual prescription torque unobtainable

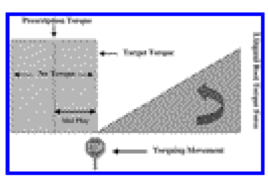

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FIGURE 5. Target torque is the final destination torque desired after the rectangular wire has delivered all its torque force and is passive. As the tooth torques toward the bracket prescription, it will stop at the target torque. The target torque is never the same as the nominal prescription torque of the bracket!

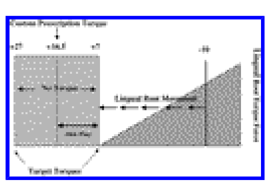

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FIGURE 6. Custom prescription torque. If a root needs to be torqued to the lingual-palatal, the Custom prescription torque equals the

target torque plus amount of slot play torque (16.5° = 7° + 9.5°). As the incisor's root torques to the lingual, moving from right (−10°) to left in the figure, the torque force will slowly diminish with tooth movement until it reaches zero at the target torque of +7°. If, however, in this example, the root was displaced excessively to the lingual more than +27°, the root would only be brought forward to +27° before all labial root movement stops (still leaving excessive labial crown torque)

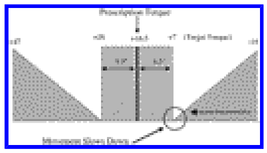

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FIGURE 7. Slowing torque movement. As a root moves toward the target torque, the torsional force diminishes until it is quite low just before the target torque is approached. Note circled area on graph

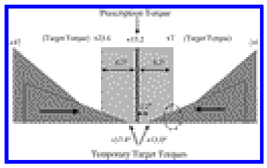

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FIGURE 8. Hypertorque graph. Temporary hypertorque; temporarily reducing the slot play will move the target torque further to the left (to +14.5°) increasing the amount of torque force available for final torquing up to the desired target torque. Using oversized NiTi and beta titanium wires can be quite an efficient straight wire–torquing technique. Using the hypertorque wire for too long may result in overtorquing past the desired target torque of +7°.

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†A + sign indicates the direction of inclination of the facial surface measured from a perpendicular to the occlusal plane with the gingival margin more to the lingual than the incisal edge (ie, palatal root torque), and a − indicates the opposite with the gingival margin being more to the labial than the incisal edge.

‡The moment of inertia of a rectangular section = (shortest side of the rectangle)^3 × (longest side of the rectangle).
