Stiffness of Incisor Segments of Edgewise Arches in Torsion and Bending

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A quantification of mean arch stiffness and force values in mass torquing of four model maxillary incisors with different Edgewise arch configurations, wire sizes and materials.

KEY WORDS • ARCHWIRE • BENDING • MECHANICS • TORQUE •

The direct contact between arch wire and bracket in the Edgewise orthodontic appliance enables the transmission of the full complement of six force system components with corresponding components of displacement — linear forces in all three planes, and rotating couples operative around all three axes in space.

The orientations of the linear forces can be identified as faciolingual, occlusogingival, and mesiodistal. The orthodontic terminology usually applied to the rotational couples is first order (axial rotation), second order (mesiodistal tipping), and third order (faciolingual tipping).

This paper addresses one of those six force-displacement sets — the third-order couple and related rotation (torque), with some related linear actions.

In the field of engineering, torque is defined as an internal force system produced within a member like an archwire by outside force couples applying a twisting action. Comparable external resultant couples will be applied when this internal force is released through a mechanism such as the Edgewise bracket slot interface with the archwire.

The term torque has two different but related meanings to the practicing orthodontist. It refers to the control or change of faciolingual tooth root angulation, and it also refers to the amount of twist applied to an arch wire in bracket engagement (activation).

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This study was undertaken to fill a need perceived by the Authors for quantification of a torsional archwire parameter that would enable structural comparisons of torquing appliances. It is focused on Edgewise mechanics intended to torque the maxillary incisor segment as a unit. Although torsional stiffnesses were specifically sought, secondary effects of third-order activation brought bending stiffnesses into consideration as well.

— Background —

Torque in the maxillary incisor segment is generated by progressive insertion of a prepared archwire, first into posterior anchorage attachments, then into the anterior brackets, with a third-order elastic deformation (twist) as it is engaged in the anterior bracket slots.

Use of rectangular wire engaging rectangular slots requires that the cross-sectional diagonal dimension of the wire be greater than the occlusogingival slot width in order to provide the engagement necessary to transmit the torquing forces to the tooth.

A smaller rectangular wire or a round wire can rotate within the bracket slots, requiring torquing spurs or other means of engagement to produce this action. The free rotation and more selective engagement of such wires alters the force system beyond the scope of this study.

Incisor-segment torquing mechanics are ordinarily undertaken to produce one of two orthodontic movements —

- **Bodily movement,** with no alteration in labiolingual inclination
- **Root tipping,** with minimal crown movement

Related Linear Forces

Both movements require a combination of an anteroposterior linear force and a third-order couple (labiolingual root torque). The couple-to-force ratios required to generate these movements are similar but not identical in magnitude (Nikolai 1974), because there are differences in the mechanical objectives of the two force components.

In both cases, the torsional and labiolingual components of the force system delivered to the incisor segment are separable within the appliance, so the torsional characteristics may be examined separately.

In bodily movement, the supplementary linear force is great enough to move the crown and therefore the entire tooth lingually while the lingually-directed apical linear component of the couple is used to move the root tip lingually as far as the crown.

In root movement, the emphasis is different; the couple produces the desired rotational (tipping) movement while the supplementary linear force functions only to prevent reciprocal labial crown displacement.

Torquing Forces

Torque in a rectangular wire powers one of the most potent orthodontic force systems, yet it is still perhaps the least understood aspect of Edgewise mechanics.

The torsional couple produced by a given amount of twist activation is directly related to torsional stiffness.

The magnitude of the third-order torque generated between wire and bracket at activation is directly related to the amount of twist applied to achieve engagement. The effective twist angle is the difference between the as-activated angular deformation of the wire and the passive angulation of the wire cross section. Orientation of the wire to the bracket

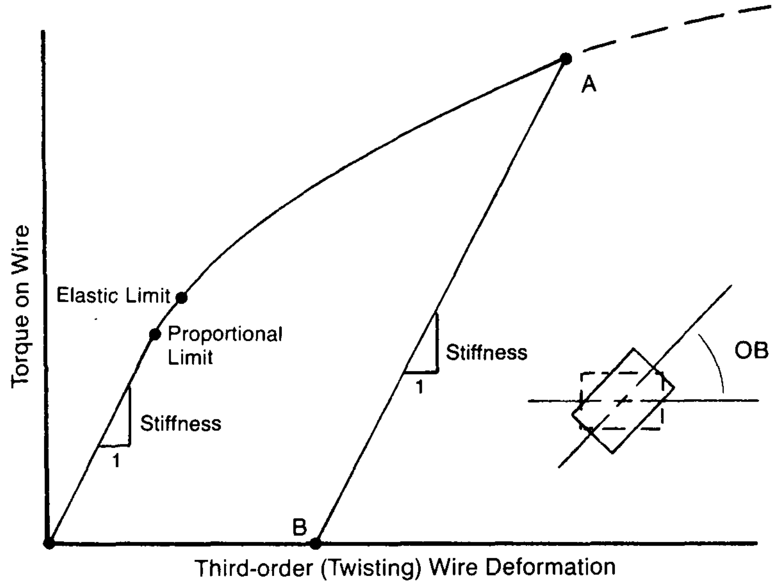

Fig. 1 **Schematic plot of Torque vs. wire deformation, as might be determined in a torsion-test apparatus. Stiffness is a measure of resistance to deformation, so it is proportional to the force required to produce a given deformation.**

This plot depicts the placement of a permanent twist in a wire, taking it beyond its elastic limit to a point A, where removal of the load produces the line AB. A permanent deformation remains (the horizontal displacement at zero load, and OB in the cross-sectional view), and a new proportional limit is established at A.

slot, the clearance between the wire and bracket, and possibly even the flexibility of the ligated bracket can affect the effective angle of twist.

Torque-twist plots are presented in Figs. 1 and 2. The deactivation plot, reflecting permanent twist changing the long-axis angulation, generally parallels the activation curve. The identical slopes of the straight-line portions of the two plots represent the torsional stiffness of the wire over the length under test.

Figure 2 indicates the initial twist activation necessary to line up the wire with the bracket slot to achieve engagement (the horizontal coordinate of point C), less a partial loss of that twist (O'B') due to clearance and bracket flexibility. Although it may be small and seem negligible in comparison with the third-order activation, this loss approaches (and may exceed) 10°. In plots such as those in Figs. 1 and 2, the actual torque (the vertical coordinate) is typically shown in units of gm-mm and the angle of twist in degrees. The third-order stiffness (the straight-segment slope) is then quantified for a given length of wire in gm-mm per degree.

The principal deformations accompanying the activation of an incisor-segment torque may be viewed from a buccal perspective (Fig. 3). In this view, the arch initially engaged into the posterior segments is a cantilever beam. At the top in Fig. 3 is the totally passive wire, adjusted

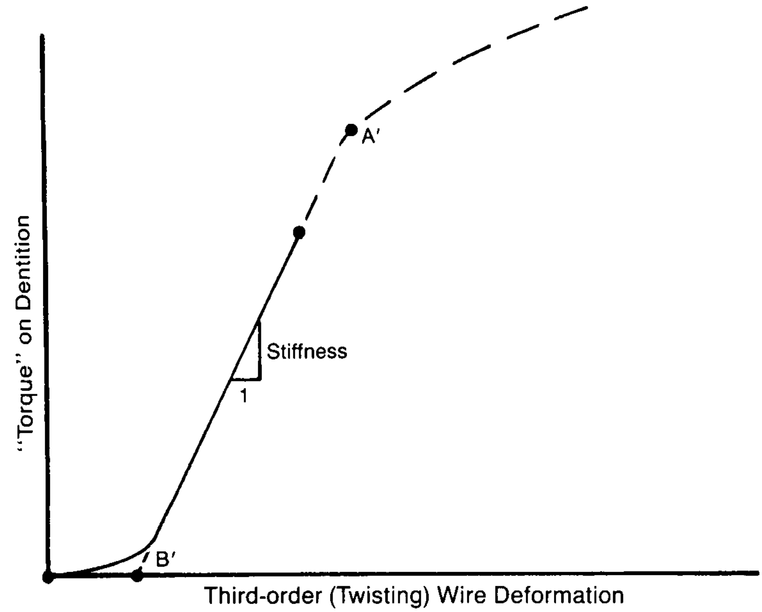

Fig. 2 Schematic plot showing the "torquing" couple applied to the dentition as the twisted wire is inserted in a bracket slot. The initial offset O'B' represents the effect of bracket clearance on initial engagement. The elevated proportional limit resulting from placing a permanent "set" in the wire (Fig. 1) is represented by A'.

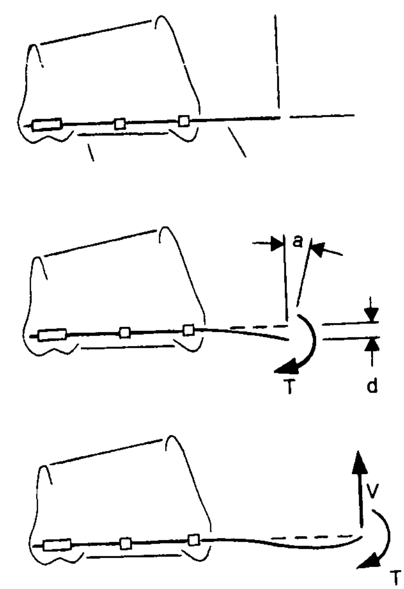

Fig. 3
Lateral views of the buccal and and incisor segments of the archwire, showing force systems and archwire deformations in the application of lingual root torque. Arrows show force systems on the wire, which are opposite to the resultant actions on the teeth.

Top — incisor section passive

Center — torque activated for bracket engagement

Bottom — wire twisted and engaged to produce lingual root torquimg action for the desired action. The light vertical and horizontal lines are references for angular and vertical displacements. The length of the archwire emerging from the posterior segment is purposely exaggerated in these sketches to magnify actual displacements.

The center sketch shows the partially activated arch. A lingual root torque has been initiated, but bracket engagement has not been accomplished. Note that accompanying the angular deformation a is a vertical deflection d; had the load been applied to produce labial root torque, the vertical deflection would have been upward.

If the anterior wire segment had been above the bracket initially, so that it became horizontally aligned with the anterior brackets with the third-order activation to generate the torsional couple, engagement could be completed with ease and no vertical components of force induced.

However, the lower sketch of Fig. 3 assumes vertical alignment of bracket and wire prior to angular deformation for activation. The displacement of magnitude d, necessary to restore alignment after the torsional activation, requires a force V. The induced force system therefore includes an extrusive force in addition to the lingual root torque.

Magnitudes of d and V have been found to be mutually proportional — the typical elastic response of a metallic member. On the basis of this relationship, the vertical stiffness of the incisor section of the wire is defined indirectly in this report, as the ratio of V to d as measured at the midline.

The published dental literature contains few articles directly addressing these clinically obvious effects. Holdaway (1956), Rausch (1959), and Burstone (1962) have commented individually from a clinical perspective on the importance of torque control and the difficulty of maintaining it in anterior-segment mechanics. Schrody (1974) simulated lingual root torquing in the laboratory, studying primarily buccal-segment reactions not addressed in the present study, and observed the vertical responses accompanying third-order activation in the anterior segment. Steyn (1977) used laboratory models to investigate the periodontal response to torquing mechanics and the division of anterior torque among central and lateral incisors.

The objective of this investigation was to quantify and compare torsional stiffnesses in the incisor segment of a sample of Edgewise arches, concurrently investigating the related vertical action in the incisor region. Procedures parallel those of Leaver and Nikolai (1978), who evaluated the Begg torquing auxiliary from a structural standpoint. That work quantified the midarch torsional and vertical response characteristics of auxiliary designs commonly employed in Stage III of classical Begg mechanics.

— Methods and Materials —

The testing apparatus employed in this investigation was designed and fabricated specifically to enable quantification of incisor segment torque in the Begg auxiliary study (Leaver and Nikolai 1978). The test arch form was based on the Bonwill-Hawley diagram, fabricated by the Senior Author to the dimensions shown in Fig. 4.

Dentition Model

Also shown in Fig. 4 is an arch engaged in the dentition model. The posterior segments are solid units, rectangular in the sketch, rigidly anchored into the base of the testing apparatus. The simulated maxillary dentition represents a level alignment with first bicuspids extracted and cuspids retracted. Nontorqued cuspid and bicuspid brackets and buccal tubes are affixed to the posterior seg-

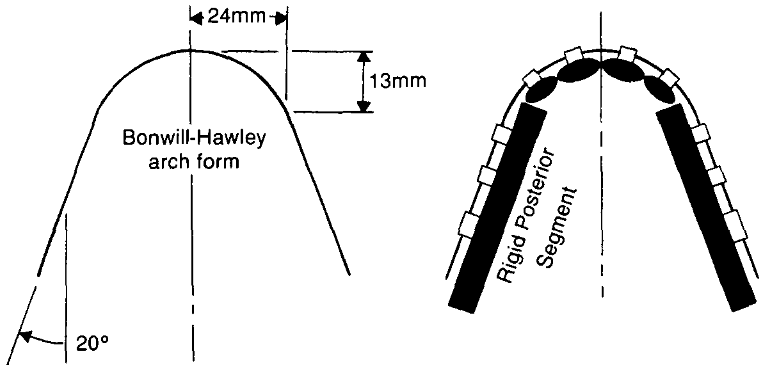

Fig. 4 **Occlusal views of the passive arch form of the test specimens and a test arch engaging the experimental dentition model**

ments, appropriately spaced. Since these attachments are all rigidly united with each other and with the base, the posterior attachments are primarily for visual orientation; they have negligible mechanical effect in this model.

The four maxillary incisors were individually fabricated from acrylic sheet, with central incisors 9mm wide and laterals 6.5mm. The space between the distal of the lateral incisor brackets and mesial of cuspid brackets was 7mm. Wide 4mm twin brackets with nontorqued slots were bonded to the model incisors. Ligation was with elastic O-rings.

Measurement Method

Figure 5 is a schematic buccal view depicting the means used for measuring the torque transferred between an archwire and the four model incisors. Test activation was accomplished from the passive state, following arch engagement, by means of a system of monofilament lines, pulleys, and weights.

Individual lines attached to the incisors at their incisal edges and apices were merged into one incisal line and one apical line, with each running over a pulley to a weight basket. The elasticity of the monofilament line tended to equalize forces among the incisors. Equal weights in the two identical baskets gave a resultant couple, transmitted from the lines through the teeth and brackets to the archwire.

The known weight and geometry enabled quantification of the torsional couple T as indicated in the sketch.

Angular position and change were measured with a fixed protractor and a pointer embedded in the apex of one of the central incisors.

A linear scale on the apparatus base adjacent to the midline was used to determine vertical position and displacement of the midpoint of the archwire. Vertical force was measured with a hand-held calibrated force gauge as the arch was physically restored to its original passive vertical position.

The midline measurements represent the sum of effects in central-lateral and lateral-cuspid spans of the archwire.

Test Arches

Test arches were prepared from straight lengths of wire using ordinary clinical procedures. Templates were used to ensure symmetry. Outside occlusogingival dimension of teardrop loops was 6mm. Each loop was fabricated with a 1mm

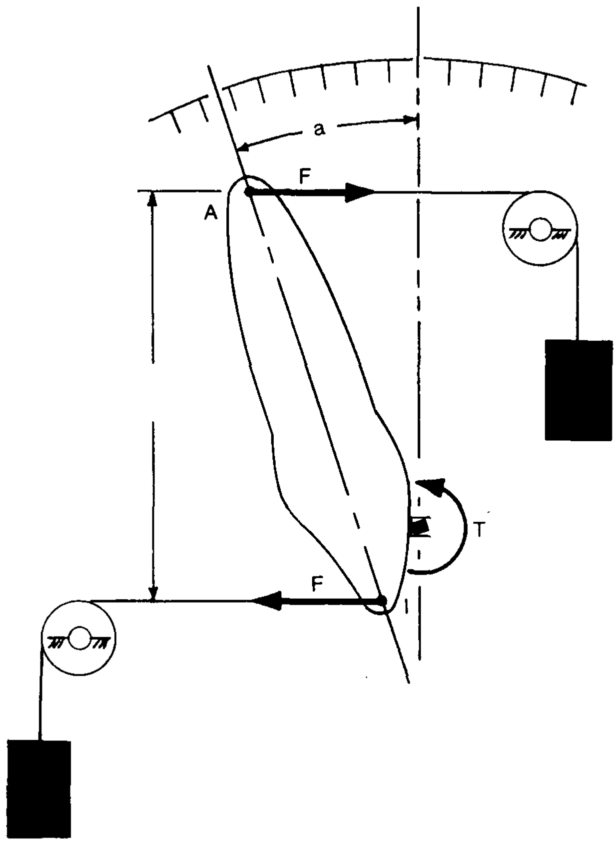

Fig. 5 **Buccal view of the torsion test configuration**

passive opening to prevent contact between legs on third-order activation.

The wire sample is shown in Table 1. Each was fabricated in four arch designs:

1. Flat arch requiring no further fabrication
2. Arch with V-bends immediately mesial to the cuspid bracket positions
3. Arch as in 2, with localized permanent twists immediately mesial to the V-bends
4. Arch with gingivally-oriented teardrop loops mesial to the cuspid brackets, and permanent twists added immediately mesial to the loops

The amount of permanent twist placed in each arch varied with on wire size, ranging between 30° and 45°. The two .022 stainless-steel wires in the three arch designs incorporating V-bends and loops were also subjected to a stress-relieving heat treatment at a temperature of 850°F for four minutes. The Elgiloy wires were not heat-treated prior to testing.

Six specimens of each arch subsample were fabricated for evaluation in the testing apparatus, so the total sample consisted of 264 specimens with no stress relief and 36 stress-relieved specimens. The order of testing of individual specimens was randomized.

Testing Procedure

In a typical test, the arch was engaged in the posterior supports of the apparatus and the four incisors ligated to the wire. Following angular and vertical reference measurements, increments of weight were added sequentially to the baskets in equal pairs, and central incisor angulations read at each addition. The arches were not unloaded between readings, but each test specimen was fully unloaded from the point of maximum activation to check for permanent set. Eight sets of coordinates were obtained from each test.

With the smaller wires, the range of angular deformations extended to the angular limit of the testing apparatus. The total twist activation exceeded 20° in nearly every test.

In addition to the torsion measurements, three sets of vertical deflection (bending) coordinates were recorded in each test.

Torsional and vertical stiffnesses were determined by linear regression. The first coordinate pair was discarded to eliminate the initial nonlinear clearance-related portion of the torque-twist curve (Fig. 2). In addition, because activations of lighter wires occasionally reached the angular displacement limit of the apparatus, the eighth coordinate pair for that subsample was not used. Accordingly, the torsional stiffness of each specimen was based on a linear regression using six or seven data points.

— Findings and Discussion —

The arch specimens were subjected to maximum incisor-segment torsional couples of 2800gm-mm to 4300gm-mm. The resulting third-order activations ranged between 20° and 45°.

No archwire tested showed measurable permanent set on unloading from the maximum activation.

Correlation coefficients obtained with regression analyses to quantify torsional stiffnesses were consistently high (rarely below 0.9).

The typical range of bracket-slot clearances, calculated on the basis of nominal bracket and wire dimensions, was between 2° and 7°.

If the stiffness is known, the torque transmitted from the arch to the incisor segment with a specific third-order activation may be estimated. It is important to note that length between load-transfer

Table I

Archwire Sample

Dimensions in Nominal Inches

Wire	.018 Slot		.022 Slot	
Rounded Edgewise Stainless Steel ○	.017 N/A	.018×.025	.019×.026	.0215×.027
Rectangular Edgewise Stainless Steel □	.017×.022	.018×.025	.019×.025	.0215×.028
Rectangular Edgewise Chrome-Cobalt ▿	.017×.022	.018×.025	.019×.026	.0215×.028

○ TP Laboratories LaPorte, IN
□ Permachrome, Unitek Corp., Monrovia, CA
▿ Blue Elgiloy, Rocky Mountain Orthodontics, Denver, CO

sites has a significant effect on stiffness, so in this case interbracket distances between cuspids and lateral incisors influence the torsional stiffness. The results presented here are associated specifically with the model arch form and mesiodistal geometry.

Ignoring the possible contribution of bracket deformation, a simple formula for the line B′C in Fig. 2 for an arch segment may be written as:

$$\text{Torque(gm-mm)} = \text{Stiffness(gm-mm/deg)} \times \text{Twist(deg)}$$

Again, the effect of "unencumbered" wire length on specific results must be noted. In torsion, the stiffness is theoretically inversely proportional to this length; in the bending mode the impact of length is hypothetically cubic. Accordingly, the tabulated vertical stiffnesses may be substantially influenced by the interbracket distances between the model lateral incisors and cuspids.

The testing verified the effect illustrated in Fig. 3 — the activation of third-order torque in the incisor segment was partially transformed into second-order (bending) couples in the wire distal to the lateral incisors. The bending response is related to arch curvature, so it is present to some degree throughout the arch, but it is especially significant between lateral incisor and cuspid. The effect at that location is a secondary vertical action with an extrusive effect on the incisors.

The expected observation of the downward displacement of the incisor section of the wire during torsional loading was the motivation for the secondary objective of the research design, which was directed toward quantification and coordination of this effect in relation to the third-order action.

Bending stiffness was computed from vertical force and displacement data. The effect of clearance between bracket and wire in this loading is negligible. Even though just three data points were obtained for each specimen, the high regression coefficients indicated a reasonable approximation of a straight-line relationship. The equation comparable in format to that for torque is:

$$\text{Force(gm)} = \text{Stiffness(gm/mm)} \times \text{Deflection(mm)}$$

An equally important relationship is that between bending stiffness and span, which has an effect inversely proportional to the *cube of the length* (length^3). This makes bending stiffness much more sensitive to tooth and bracket dimensions than torque, so even small changes in the span from lateral to cuspid can lead to dramatic changes in the relation between the clinical effects of bending and torsion.

Although data analyses can reveal whether or not differences in stiffnesses are statistically significant, they are not necessarily indicative of significance in the clinical setting. Accordingly, guideline figures were derived on the hypotheses that 25gm of force and 1mm of crown or root movement are clinically significant. For incisors of average size, assuming the center of resistance to be midway between the apex and cemento-enamel junction, these hypotheses suggest that the following differences would be clinically significant for the incisor segment:

500gm-mm in torsional couples
25gm-mm/deg in torsional stiffness
50gm in vertical forces
20gm/mm in vertical stiffness

Torsional Stiffness

The three-way analysis of variance performed on the torsional stiffnesses of the sample nucleus indicated that more than 70% of the variance was attributable to wire size and arch design, with wire size somewhat more dominant than design. Differences among wire materials was almost nil.

Because rectangular shaft theory indicates that torsional stiffness is proportional to the *fourth power* of cross-sectional area (Popov 1968), the finding of significant differences among wire sizes was expected. Furthermore, the lack of significant differences between subsamples partitioned by arch material was not surprising, inasmuch as stainless steel and Elgiloy exhibit very similar stiffnesses (elastic moduli).

The effects of arch design reflected the effects of loops and bends on overall length of wire. In the order of increasing torsional flexibility, the plain arch form was the stiffest, followed by the arches containing V-bends only, V-bends and permanent twists, and loops and permanent twists. All subsamples of arch design were significantly different statistically, and most designs differed "clinically" from each other. Only the flat arch and that with V-bends only showed no "clinical" difference.

Table 2 presents the mean torsional stiffnesses for the 44 arch subsamples. The decreases in torsional stiffness with the addition of bends, twists, and loops, are evident in the table. Noteworthy exceptions are the three reversals compared to main-effect rankings between plain and V-bend arches of .018×.025 and .0215×.027 TP wires and the .017×.022 Elgiloy wire.

Vertical Bending Stiffness

The results of the vertical stiffness data analyses paralleled those of torsional stiffness. The induced vertical force (V in Fig. 3) was found to be directly proportional to the unrestrained vertical deflection of the incisor section of the arch. The force-deflection ratios for the test subsamples are the vertical stiffnesses in Table 3.

Of the three independent variables investigated, only arch material was found to have an insignificant effect on vertical stiffness. Wire size was found to be most significant, in keeping with structural theory which indicates that stiffness in bending is greatly affected by cross-sectional dimension in the direction of deformation.

Most differences in wire size and arch design were judged clinically significant by the criteria established earlier.

Table 3 shows the mean vertical stiffnesses for the 44 subsamples. The general patterns of increasing stiffness with

Table 2

Mean Torsional Stiffness in gm-mm per degree

Wire	Size	Flat	V-Bend only	V-Bend & Twist	Loop & Twist
Round	.018×.025	167	189	126	116
Edge	.019×.026	281	239	168	118
Steel	.0215×.027	246	253	204	166
Rect	.017×.022	146	138	119	74
Edge	.018×.025	185	168	141	112
Steel	.019×.025	273	249	163	119
	.0215×.028	300	276	252	199
Rect	.017×.022	146	182	151	89
Edge	.018×.025	178	168	154	124
Cr-Co	.019×.026	258	203	149	126
	.0215×.028	287	257	237	169

wire size and increasing flexibility with length of wire are evident. The analysis-of-variance summary indicated significant three-way interaction within the set of means that did not occur in the torsional stiffness data analysis. Several individual reversals in rank are evident with variation in arch design for specific wires. In this analysis the effect of V-bends was less strong than that which appeared in the torsional stiffness data, where the associated geometry is more significant.

For all wires the inclusion of loops resulted in clinically significant reductions in vertical stiffness.

Stress Relief

The stress-relieved subsample was limited to arches formed in stainless steel wires of largest cross sections, including the three designs incorporating second-order bends and/or twists. Analyses of variance were used to compare these specimens with their untreated counterparts.

The main-effect results showed no significant influence of stress relief on torsional stiffness.

Table 4 presents the mean torsional and vertical stiffnesses for the stress-relieved specimens and comparable arches of the nucleus. The table reflects mixed effects of stress relief. Lane and Nikolai (1980) reported increases in the stiffnesses of retraction loops after stress relief when activated mesiodistally. Opposite findings in some wires in the present study in which looped arches received an activation which was generally directed occlusogingivally indicate the complexity of influences on loop stiffness.

No nonelastic behavior (permanent deformation) was noted in the stress-relieved specimens. Because the results of this portion of the study showed insignificant and inconsistent changes in stiffness, stress relief of these arch designs following fabrication for the purpose of increasing stiffness seems unwarranted.

Preactivation Calculations and Clinical Implications

Twist angle and torquing key force are the most common criteria for the measurement of torsional activation, with clinicians relying principally on treatment experience in selecting magnitudes.

Table 3

Mean Vertical (Bending) Stiffness in Grams per Millimeter

Wire	Size	Flat	V-Bend only	V-Bend & Twist	Loop & Twist
Round	.018×.025	122	133	112	68
Edge	.019×.026	175	152	150	100
Steel	.0215×.027	175	148	190	128
Rect	.017×.022	70	94	91	55
Edge	.018×.025	107	115	110	70
Steel	.019×.025	187	146	154	92
	.0215×.028	194	203	179	144
Rect	.017×.022	84	86	75	59
Edge	.018×.025	129	108	117	85
Cr-Co	.019×.026	147	136	123	87
	.0215×.028	213	208	193	158

The results of research such as this, together with biomechanically-based quantified force systems for specific tooth movements, could enable computation of reasonable approximations for appropriate third-order activations.

Clearly, the present research has examined a very restricted sample of model arches and torque application designs. "Artistic" arch details are absent, and no variations in arch form or dimensions were included in the research design. Furthermore, the parameters of interest were evaluated only at the initiation of lingual incisor segment movement, with sizeable interbracket distances between cuspids and laterals. Consequently, the quantified torsional and vertical stiffnesses provide only a limited set of values at the lower end of the scale that would be encountered in clinical application.

A rationale and development of hypothetical physiologically-based orthodontic force systems have been presented by Nikolai (1975) and Wagner (1981). They suggest values of 150gm for the linear force and 1700gm-mm for the counter-tipping couple for "continuous" bodily movement of an average maxillary incisor segment.

Using Table 2 and the torque formula presented previously, the approximate initial input to an .018×.025 retraction arch with loops would be about 18°. The hypothesis yields approximately 3200gm-mm of initial torque and 230gm initial holding force for root tipping of an average segment.

It has been previously shown that with the wires and incisor bracket slots at the same vertical level after arch insertion into the posterior attachments, activation of lingual root torque for incisor bracket engagement results in a downard force on the incisor segment. The clinical effect of this resultant downward force will be

Table 4

Mean Stiffness Values
for .022" Stainless Steel Archwires
With and Without Stress Relief Heat Treatment

Wire	Stress Relief	V-bend Only	V-bend & Twist	Loop & Twist
Torsional Stiffness in gm-mm per Degree				
Round Edge Steel	No	253	204	166
	Yes	225	228	166
Rect Edge Steel	No	276	252	199
	Yes	333	265	192
Vertical (Bending) Stiffness in Grams per Millimeter				
Round Edge Steel	No	148	190	128
	Yes	195	229	146
Rect Edge Steel	No	203	179	144
	Yes	233	204	140

extrusive displacement. Using Table 3, adjusting an .018×.025 arch about 2mm above the central incisor bracket slot could counteract this potential extrusion.

Table 5 presents changes in vertical force against the incisor segment with changes in long-axis angulation for the non-heat-treated sample nucleus. Compensating for clearances, products of Table 5 values and third-order activations yield the vertical-force magnitudes for the test arches.

These values are dependent on overall arch geometry as previously mentioned, so they are not universally applicable. The trends of vertical force potential, increasing with wire size and decreasing with length of wire in the anterior segment are, however, noteworthy.

Table 5

Differential in Induced Vertical Force
per Unit Change in Long axis Angulation

(Grams/Degree)

Wire	Size	Flat	V-Bend only	V-Bend & Twist	Loop & Twist
Round	.018×.025	14.7	15.4	10.1	9.1
Edge	.019×.026	9.5	12.7	12.7	10.3
Steel	.0215×.027	19.7	14.6	14.3	14.8
Rect	.017×.022	7.8	9.9	6.6	6.4
Edge	.018×.025	10.9	12.5	9.3	8.1
Steel	.019×.025	12.0	12.1	13.1	10.0
	.0215×.028	20.9	18.3	15.7	15.5
Rect	.017×.022	8.6	7.7	6.7	7.1
Edge	.018×.025	14.6	11.8	9.4	10.9
Cr-Co	.019×.026	11.2	11.3	9.9	10.1
	.0215×.028	22.7	23.8	18.9	15.7

— Summary and Conclusions —

A mechanical model dentition is used in a laboratory study to relate incisor segment third-order activation to actual torque induced upon engagement of maxillary Edgewise arches. A portion of the sample of stainless-steel aches was subjected to stress-relief heat treatment. Torsional stiffness values are calculated, and the accompanying vertical displacement force on the incisor segment is evaluated.

The results seem to warrant the following conclusions.

• Broad ranges of incisor segment torque magnitudes may be obtained from rectangular orthodontic wires and arch designs presently in common clinical use.

• Torsional behavior is associated with the elastic shear modulus or modulus of rigidity, which is essentially the same for all stainless and chrome-cobalt alloys.

• Vertical extrusive force is generated as a secondary effect directly related to the torsional stiffness and torque activation.

Compensation is possible through archwire adjustments to cause the wire to lie above the bracket slots of the incisor segment before activation of lingual root torque.

• Stress relief of rectangular stainless steel arches following placement of V-bends, twists, or loops did not have a significant effect on force values.

• Further investigation, modeling other configurations such as labiolingual movements of the six maxillary anterior teeth to determine the torsional stiffnesses for commonly-used arches could be worthwhile. Similarly, the quantification of torsional stiffnesses of arches fabricated in rectangular nickel-titanium, titanium-molybdenum, and braided stainless-steel wires may be of value. A/O

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