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Torque capabilities of self-ligating and conventional brackets under the effect of bracket width and free wire length

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Structured Abstract

Objectives – To numerically investigate the torque capacity of conventional and self-ligating brackets under the effect of varying bracket width and free wire length.

Material and Methods – Finite element models of three kinds of orthodontic brackets in the 0.022-inch slot size were investigated: Discovery, Damon 3MX, Speed. Additionally, finite element (FE) models of Speed and Damon brackets were generated with the same width as the Discovery. From the left upper incisor to the right upper canine, four brackets each were modelled. The total wire length at the upper right incisor was kept constant at 12 mm for all brackets types. For the Discovery brackets, the wire length was increased from 12 to 16 mm in 2-mm steps. A torque of 20° was applied to the upper right incisor with 0.46×0.64 mm^2 ($0.018'' \times 0.025''$) and 0.48×0.64 mm^2 ($0.019'' \times 0.025''$) wires. Wires made of stainless steel, titanium molybdenum and nickel titanium were studied. Torque angle/moment characteristics were recorded.

Results – Wider brackets showed more torque control capability (e.g. Discovery: 10.6 Nmm, Damon: 9.2 Nmm, Speed: 4.0 Nmm for the NiTi wire). Even with the same width as the Discovery bracket, Damon and Speed brackets showed lower torque capability than the Discovery bracket. Increasing the free wire length decreased the torsional stiffness of the wire and thus decreased the torque capability.

Conclusion – The results showed that the bracket design has less influence on the torquing moment than other parameters, such as bracket width, free wire length, wire/slot play or misalignment.

Key words: biomechanics; brackets; finite element methods; self-ligation; torque

Introduction

In orthodontics, torque is employed to alter the inclination of teeth, particularly the incisors. The interaction between the bracket of an axially rotated tooth and arch wire produces a torquing moment. In general, the extent of change in the buccolingual inclination of crowns depends on the wire torque stiffness, the bracket design, the wire/slot play and the mode of ligation (1–3). An additional factor in the clinical routine application of torquing moments relates to the interbracket distance and the bracket width (4–6). The wide array of combinations of the altering factors in defining torquing moments makes the empirical clinical determination of the appropriate torquing method a difficult task for the practitioner. Currently, there is a lack of evidence regarding the torque characteristics of various bracket/arch wire combinations (3, 7–10). This may be attributed to the complexity of the experimental configuration required in laboratory studies (11), and the multiplicity of factors needed to be controlled in a clinical setting, including individual response to moments applied, variability in malocclusion and the potential effect of other auxiliaries or treatment utilities, affecting torque.

In the past two decades, there has been a boost in the manufacturing and release of self-ligating appliances with active or passive ligation modes, which attract more and more eyesight because of the time-saving ligation mode and the potential alterations in the load and moment expression during mechanotherapy. Whereas some of these systems seem to present reduced friction *in vitro*, their torquing characteristics remain uncertain (8).

In an earlier study it could be demonstrated that self-ligating Damon and Speed brackets had lower torque capabilities compared with conventional Discovery brackets (12). The torsional stiffness of the arch wire was the significant contributing factor with regard to the torquing moment. However, as the self-ligating brackets were narrower than the Discovery bracket, it was difficult to attribute a better torque capability to the Discovery bracket compared with self-ligating brackets. This is due to the fact that the bracket width influences the free wire length and although the torsional stiffness of an arch wire and thus the moment/torque characteristic is dominated by the wire cross-section, changing the free wire length is another method to control torque expression.

The objectives of the present finite element method (FEM) study were:

1. To analyse the overall bracket effect, that is, bracket width, wire/slot play and ligation mode, on the torque capability of conventional and self-ligating brackets.
2. To eliminate the influence of bracket width by rescaling self-ligating brackets to the width of a conventional bracket and thus allowing the analysis of the ligation method effect on the torque capability in the same bracket width.
3. To analyse the torque capability of a conventionally ligating bracket under different free wire lengths by varying the interbracket distance of the upper incisor brackets.

Materials and methods

Three types of bracket systems were selected for this study: self-ligating Hanson Speed™ (Strite Industries, Cambridge, ON, Canada) and Damon™ 3 MX (Ormco, Glendora, CA, USA) brackets, as well as the conventionally ligating bracket Discovery® (Dentaurum, Ispringen, Germany). The widths of the brackets were as follows: Discovery 3.36 mm, Speed 2.33 mm and Damon 2.75 mm. FE models of the brackets were generated with the respective dimensions. Three-dimensional models of the Speed and Damon brackets were generated on the basis of cross-sectional μCT-scan views of the different specimens (Skyscan 1072HR, Aartselaar, Belgium). The 3D reconstructions were performed using the especially designed software ADOR-3D (13). The CAD data of the conventional Discovery brackets were provided by the manufacturer (Dentaurum, Ispringen, Germany). Additionally, two further FE models were generated for the Damon and Speed brackets by rescaling the geometries such that they had the same width as the Discovery bracket. All brackets had a 0.022-inch slot size. From the left upper incisor to the right upper canine, four brackets were included in the numerical model. Figure 1 shows the FE models of the brackets with varying bracket widths and Fig. 2 the full model with four brackets and the engaged arch wire.

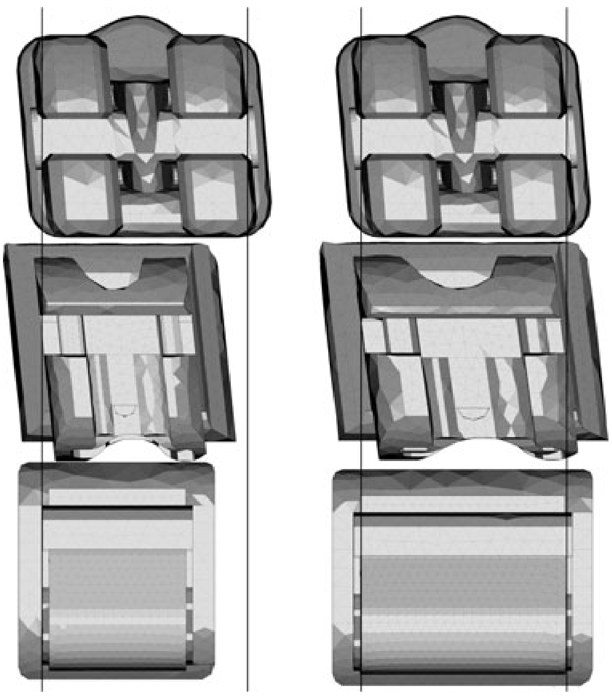

Fig. 1. Finite element (FE) models of the brackets investigated: the left column shows the Damon and Speed brackets with original widths compared with the Discovery. In the right column, brackets are displayed with the width of the Discovery bracket.

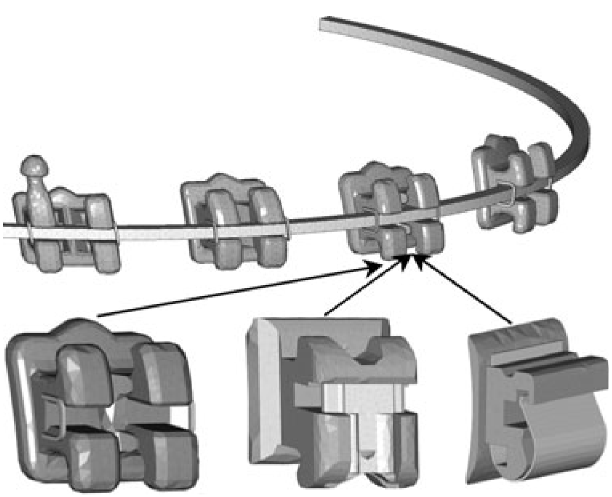

Fig. 2. Finite element (FE) model of the overall configuration. Four brackets of each bracket type from the left upper incisor to the right upper canine have been integrated in the FE model.

The following material parameters were assumed for the brackets and ligatures: Discovery – steel (Young's modulus 200 GPa, Poisson's ratio 0.3) with conventional elastic ligature (0.1 GPa, 0.3), Damon – steel (200 GPa, 0.3) with passive steel clip (200 GPa, 0.3), Speed – steel (200 GPa, 0.3) with active nickel titanium clip (superelastic behaviour, 0.3).

The total wire length from the left upper incisor bracket to the right lateral incisor bracket was kept constant for all five bracket/wire configurations at 12 mm (see Fig. 3) to determine the effect of the bracket type, including ligation mode and bracket width. To determine the influence of a varying free wire length, the total wire length was increased from 12 to 16 mm in steps of 2 mm in the model with the Discovery brackets (Fig. 3), that is, the free wire length was increased from 8.64 to 12.64 mm. The following arch wires were used with dimensions of 0.46×0.64 mm^2 ($0.018'' \times 0.025''$) and 0.48×0.64 mm^2 ($0.019'' \times 0.025''$): stainless steel (short code in figures SS, Young's modulus 200 GPa, Poisson's ratio 0.3), titanium molybdenum (TMA, 80 GPa, 0.3) and nickel titanium (NiTi, superelastic behaviour, 0.3).

The torque capabilities of different bracket/arch wire combinations were determined as follows: A torque of 20° was applied to the upper right incisor by rotating its bracket by 20° in steps of 0.5° along the central axis of the slot, corresponding to the wire axis. All other brackets remained fixed in all three planes of space. A free mobility of the wire within the bracket slot was given by performing so-called contact analyses with a frictional coefficient μ between the bracket and the wire of 0.2. The wire mobility was thus restricted only by the slot walls and the ligature wire or clip. The torquing moment was generated by a deformation of the wire, comparable to the clinical situation.

Simulations of the torque movement were performed with the FE program system MSC. Marc/Mentat 2005. As an output, torque moment and angle values in the simulated movement were recorded by the FE software package. The torque moment/torque angle curves and maximum torquing moments at 20° torque angle were used to characterize the torquing capabilities of the different bracket/wire combinations.

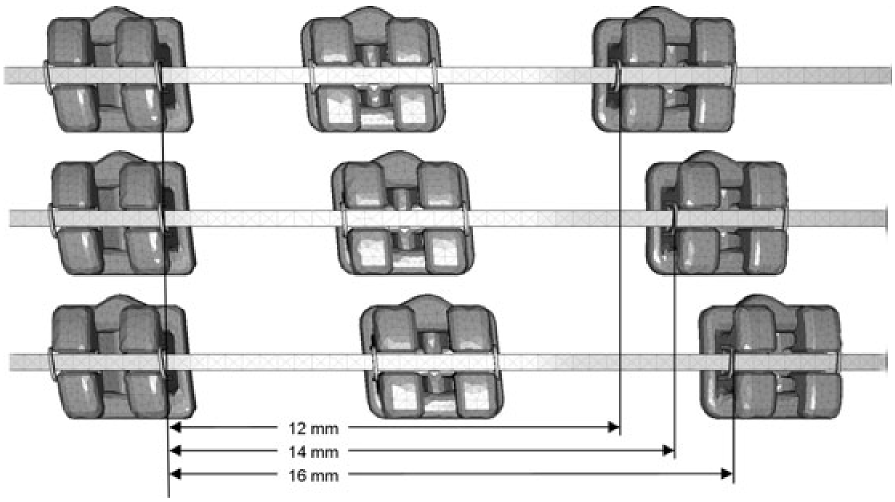

Fig. 3. Finite element model with Discovery brackets and different free wire lengths.

Results

Figure 4 shows the moment-torque activation curves of the different brackets in combination with the smaller-sized wires (0.46×0.64 mm^2, $0.018'' \times 0.025''$). The following facts can be derived from the graphs. Obviously, all curves have three segments, separated by two bends. The first bend represents the play of the wire in the slot of the torqued bracket of the upper right incisor; up to this point, no moment is generated. For the Discovery bracket, this bend can be found at a torque angle of 9.0° (Fig. 4A). A second bend is visible when the torqued wire comes into contact with the slot of the neighbouring brackets (18.5°). For all configurations, the fourth bracket in the model did not show an effect on the activation curves, and there was no further bending in the curves. The influence of the wire material is clearly visible, and the steel wire generates the highest torquing moment and steepest increase, followed by the TMA and the NiTi wires.

For the other two brackets, we have to distinguish between the activation curves generated by the brackets with commercial width and by the widened brackets. The wide Damon brackets display curves that are quite similar to the curves generated by the Discovery bracket in combination with the respective wire (Fig. 4B). Bends can be identified at 8.8° and 18.7°. However, the Damon with the smaller, that is, conventional width, seems to have slightly increased play, as bends in the curves are located at 9.8° and 19.5°. The curve of the Speed bracket (Fig. 4C) shows the characteristics of the superelastic NiTi clip. Owing to the active clip, there is almost no play visible in the curves; however, the movement of the clip allows the torqued wire to move within the bracket slot. On the other hand, the moments are extremely small and the bends visible in the curves result from the wires and activated clips that come into contact with the slot walls of the torqued bracket. The play in the neighbouring brackets with the respective bend is not visible and thus the play in the Speed without the active clip would be as high as 16.5°. The order of sequence of the different wires remains unchanged compared with the Discovery bracket.

The maximum torquing moments of the different bracket/wire combinations at a torque angle of 20° are displayed in Fig. 5 for the three wire types with the cross-sections 0.46×0.64 mm^2 ($0.018'' \times 0.025''$) (A) and 0.48×0.64 mm^2 ($0.019'' \times 0.025''$) (B). The bar graphs demonstrate quite clearly the effect of the wire type, the wire cross-section and the bracket width. The most effective way to alter moments and forces is changing the wire alloy or cross-section, while the bracket type is of less importance, except for the Speed. The wire sequence is clearly visible in all bar graphs of Fig. 5a, b. The highest moment of 75 Nmm is generated by the 0.48×0.64 mm^2 ($0.019'' \times 0.025''$) steel wire in combination with the conventional bracket. The Damon bracket generates roughly 20% lower moments with all respective wires, and the moments generated by

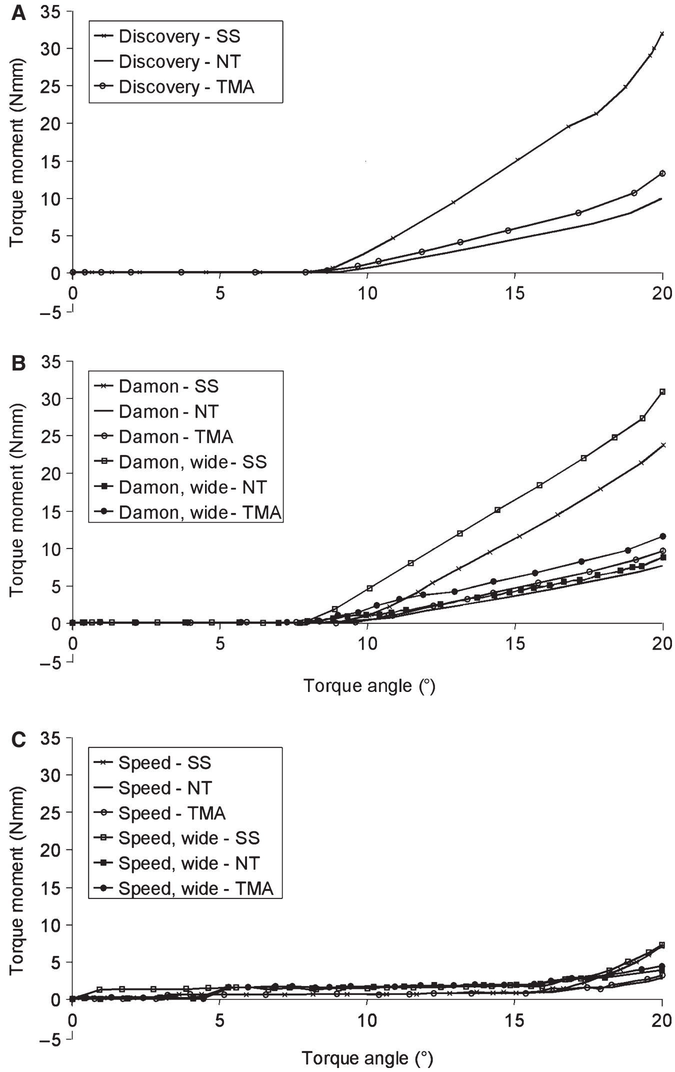

Fig. 4. Torque moment/torque angle curves of the bracket types investigated combined with the $0.46 \times 0.64\ mm^2$ (0.018″ × 0.025″) wires (SS, stainless steel; TMA, titanium molybdenum; NT, nickel titanium). (A) Discovery bracket with an elastic ligature, (B) Damon bracket with normal width and with the width of the Discovery bracket (wide), and (C) normal and wide Speed bracket. The effect of the bracket width is obvious.

the Speed bracket are only ¼ of the moments generated by the Discovery. The effect of the bracket width is best seen for the Damon bracket in combination with the $0.46 \times 0.64\ mm^2$ (0.018″ × 0.025″) wires (Fig. 5a). The moments for the Damon with the bracket width of the Discovery are increased and almost reach the values generated by the Discovery. The remaining difference may be attributed to the higher play as can be seen in the curves of Fig. 4.

Figure 6 shows the maximum torquing moments for the FE model with Discovery brackets and the $0.46 \times 0.64\ mm^2$ (0.018″ × 0.025″) wires at different free wire lengths. Increasing the wire length from 12 to 14 and 16 mm causes the decrease of the maximum torquing moment from 34 to 29 and 24 Nmm, that is, by 15 and 30%, respectively. The wire sequence remains unchanged.

Discussion

The FEM is a numerical technique used to analyse structural stress and strain states and has already been applied widely in oral biomechanical research. Owing to its flexibility, the FEM can simulate a variety of clinical situations ‘in silico’. A key element in finite element analyses is the model generation using predefined volume, planar or beam elements to discrete the structure. The accuracy of the FE results critically depends on the accuracy of the generated mesh and the appropriateness of the elements used to discrete the structure. Typically, the predictability of FE simulations is assumed to be of the order of 20%, that is, simulation results might differ up to 20% from the real situation, owing to idealization effects. This can be seen to be

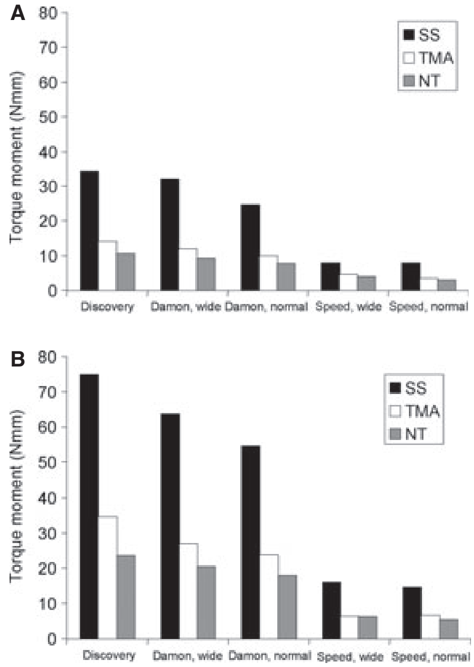

Fig. 5. Maximum torquing moment of the different bracket types and the width modifications. (A) Torquing moments at a torque angle of 20° with 0.46×0.64 mm^2 (0.018″ × 0.025″) wires. (B) Torquing moments at a torque angle of 20° with 0.48×0.64 mm^2 (0.019″ × 0.025″) wires.

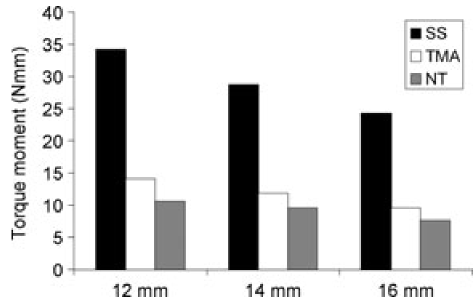

Fig. 6. Maximum torquing moment in the Finite element model with the Discovery bracket and the different free wire lengths.

adequate for the comparison of different orthodontic appliances.

The presented analysis of torque from an ideal arch delivered a couple of practicable guidelines to be used in clinical practice. However, the findings of this study should be viewed in the light of the simplifications and assumptions made. In an earlier numerical study (12), it was already indicated that ligation methods of the various self-ligating brackets can strongly influence the torquing moment. Compared with a conventional bracket, the self-ligating brackets displayed higher torque play and thus reduced torque capability. However, self-ligating and conventional brackets had different widths, which in turn is a contributing factor to the torque capabilities.

The results of this study show that the self-ligating brackets with increased bracket width increase the torquing moments by about 15% compared with originally sized brackets. Overall, the results show similar moment/torque curves, which seem to be dominated by the characteristics of the wire and the wire/slot play. Insofar, the efficacy of specialized torquing appliances, such as reverse curve of Spee NiTi wires or similar appliances, will be similar in combination with the three kinds of brackets. The wire part that is placed into the bracket slot cannot contribute to the active part of the wire. In essence, the bracket is a dead spot that functions like an annealed area (4, 6, 14–16). Increasing the width of a bracket, on the one hand, increases its control capabilities. On the other hand, however, the torquing moment is increased by the reduced wire length. Clinically, a smaller bracket is more comfortable to the patient, and a compromise must be found between bracket control and generated forces and moments in first- and second-order bending.

Further discussion of this aspect requires differentiation between Damon (or passive self-ligation in general) and Speed brackets (or active self-ligation). The behaviour of the Damon bracket compared with the conventional bracket can be characterized and explained as follows. The change of the width did not change the shape but the slope of the moment/torque curves (Fig. 4), and by this influenced the maximum torquing moments (Fig. 5). This effect can be attributed completely to a reduction in the free wire length of the wider Damon. Remaining differences between the standard bracket Discovery and the Damon bracket may be ascribed to different torque play in the systems between the wire, the slot walls and the ligature. Insofar, this result confirms that the characteristics of the moment/torque curves are dominated by the torsional stiffness of the arch wires.

The situation is completely different for the Speed bracket. The activation curves (Fig. 4c) clearly show the characteristics of the active NiTi clip. On the one hand there is almost no play, as the clip presses the wire onto the bottom of the slot and generation of the torquing moment starts with the very beginning of the torque movement. On the other hand, however, the generated moments seem to be below or close to the minimum effective values for torquing moments listed in the literature (5 Nmm, see e.g. 17, 18). Only the thicker wires or the steel wire with the smaller cross-section generate a sufficiently high maximum torquing moment. The behaviour can easily be explained by the common deformation of the superelastic NiTi clip and the wires engaged into the bracket slot that result in a relatively flat activation curve of the wires compared with the standard or passive self-ligating brackets.

Totalling, with identical bracket width, self-ligating brackets expressed less torque than the conventional bracket. This could not be attributed solely to varying wire/slot play, and thus different ligation effects have to be taken into account as well. While the effect is minor for the Damon bracket and may be attributed by the play between the wire and the closing mechanism, that is, the self-ligating clip, the effect is most obvious with the Speed bracket and the active NiTi clip in combination with the steel wire. On the one hand, the active clip results in almost zero torque loss and the activation of the wire starts immediately with rotation of the bracket, but on the other hand at the same time, maximum torque transmission is reduced significantly.

A further contributing factor to torque control is the free wire length (6, 19–21). As Figure 6 shows, stiffness in torsion and thus the generated torquing moment increases as the inter-bracket distance decreases. This result is in agreement with former experimental investigations and with theoretical considerations from basic elasticity theory. Clinically, this means that the wider the dental arch, the greater the inter-bracket distance and the lower the moment. Same considerations hold for upper and lower dental arch lengths or for changing from labial to lingual bracket systems. The influence of lingual systems on force systems will be presented in a forthcoming paper.

Further clinical factors influencing torquing moments that could not be studied are the accuracy of vertical bracket positioning or the morphology of the teeth. Several studies proposed different amounts of torque variation. Meyer and Nelson (22) described that a vertical shift of 3 mm can change the torque angle by around 15°, while Miethke (23) proposed that a torque variation of 10–15° may already arise from a vertical inaccurate placement of 1 mm. The morphology of the teeth can vary greatly and thus affect the clinical use of a torque (24). The angle between the longitudinal axis of the root and the crown at an upper central incisor can also vary (25), modifying the outcome of application of same moment on different shaped crowns. Nevertheless, the presented systematic numerical study allows to draw a couple of clinically relevant conclusions.

Conclusions

The results show that the bracket design has less influence on the generated torquing moment than other parameters, such as wire type, bracket width, free wire length, wire/slot play or degree of misalignment:

- The most important factor with respect to moment generation is the wire type.
- Wider brackets have more torque control ability and generate higher moments.
- Compared with the Discovery bracket, even having the same widths, the Damon brackets showed lower torque capability owing to higher play.
- The characteristic of the active Speed bracket in torque is dominated by the NiTi clip. Even with the same width as the Discovery bracket, it generated the lowest torquing moment.
- The increase of the free wire length decreased the torsional stiffness of the arch wire, which in turn is the main contributing factor to the torque capability.
- The larger the free wire length, the less the torquing moment.

Clinical relevance

The number of different bracket systems with varying closing mechanisms on the market is enormous. Self-ligating brackets were introduced to replace conventional ligation methods with elastics or steel ligatures and improve clinical efficacy. They are often attributed with characteristics such as 'low friction', 'low force' and 'optimized torque control'. Understanding the different mechanisms that can influence torque characteristics of different wire/bracket combinations is of major importance to estimate the outcome of a planned treatment. Aside from bracket geometry, and especially bracket width, other clinically relevant parameters can influence the torquing capacity of a given wire/bracket combination, such as wire/slot play, mode of ligation and free wire length.

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