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Torque capacity of metal and polycarbonate brackets with and without a metal slot

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SUMMARY The aim of the present study was to investigate slot deformation and the equivalent torque capacity of polycarbonate brackets with and without a metal slot in comparison with those of a metal bracket. For this purpose, the expansion characteristics and, in a further investigation, the labial crown torque of an upper central incisor, were measured in a simulated intra-oral clinical situation, using the orthodontic measuring and simulation system (OMSS). Three types of bracket with a 0.018 inch slot were tested: polycarbonate Brillant® without a metal slot, Elegance® with a metal slot and the metal bracket, Mini-Mono®. For testing purposes the brackets were torqued with 0.016 × 0.022 inch (0.41 × 0.56 mm) and 0.018 × 0.022 inch (0.46 × 0.56 mm) ideal stainless steel archwires.

In the activating experiments, significantly higher torque losses and lower torquing moments were registered with both rectangular archwires with the polycarbonate brackets than with the metal bracket. In the simulation tests, significantly higher torquing moments were registered with the metal bracket than with the polycarbonate brackets. The values for the Elegance® bracket were between those of the Mini-Mono® and Brillant® brackets. The OMSS model approximates the clinical situation, with the torque loss being notably higher than in the *in vitro* activating experiments. This is due to the adjacent teeth giving the archwire additional play. In addition, the torquing process may twist the archwire, resulting in subsidiary forces.

On the basis of the present results, all three brackets can be recommended for torquing. However, in view of the high torque losses, the torques programmed in the straightwire technique must be seen as questionable. Data should be provided by the manufacturer on the bending to be expected in polycarbonate brackets, which has to be offset by additional torque, or the bracket torque should be omitted from the technical specifications.

Introduction

According to Rauch (1959), the term ‘torque’ has two different but related meanings for the orthodontist. On the one hand it refers to buccopalatal root inclination, which can be measured on the lateral headfilm as the incisor inclination to the anterior cranial base or to the maxillary plane, while on the other it describes the activation generated by torsion of an archwire in the bracket slot. Depending on the magnitude of the torsion, the size and quality of the wire, the ‘play’ available for the wire in the bracket slot, the inclination, and the deformability of the bracket, the archwire moves the root in a palatal direction through the torsional tension induced in the activated state. Incisor torque is often the precondition for a normal interincisal angle, good incisor contact, and sagittal adjustment of the dentition aimed at optimal intercuspation.

In cases where torquing forces are required to move the apices of the upper incisor teeth palatally it has been specified by some authors that a minimum torque of 0.5 Ncm for an upper central incisor is necessary (Reitan, 1964; Jarabak and Fizzel, 1972; Moyers, 1973). Others state that an initial torque of 1.0–2.0 Ncm is required to maintain an adequate degree of torque (Reitan, 1957; Burstone, 1966; Bantleon and Droschl, 1988; Feldner *et al.*, 1994). In addition to these specified reference values, it should be borne in mind that due to the higher blood supply and more spongious alveolar bone in children, higher forces can be applied in children and adolescents than in adults.

Only a small number of studies to date have dealt with the extent of torque-induced deformation of polycarbonate brackets and with their torque capacity (Dobrin *et al.*, 1975; Feldner *et al.*, 1994; McKnight *et al.*, 1994; Alkire *et al.*, 1997). Feldner *et al.* (1994) and Alkire *et al.* (1997) investigated the torque capacity of five different polycarbonate brackets. Both reported a significantly lower torque and more pronounced deformation with reinforced and non-reinforced polycarbonate brackets than with metal and ceramic brackets. These attachments proved inefficient in view of the precise torque movement required. Only a metal slot-reinforced polycarbonate bracket displayed adequate stability in both investigations and could thus be used for torquing. Dobrin *et al.* (1975) also reported that two different non-reinforced polycarbonate brackets displayed pronounced deformation with increasing torque. Attaining a working range of 2.0 Ncm would involve disproportionate bending of the bracket.

Another influencing factor, the interbracket distance, is determined by tooth and bracket width (McKnight *et al.*, 1994). In the literature it is stated to be approximately 6 mm in the upper incisor region (Holt *et al.*, 1991; Feldner *et al.*, 1994). The factor most strongly influencing torque movement is the vertical positioning of the bracket on the labial surface (Dellinger, 1978; Ferguson, 1990; Muchitsch *et al.*, 1990), although data on the extent are rarely given. Meyer and Nelson (1978) reported that a bracket offset vertically by 3 mm on a lower premolar changes the torque angle by up to 15 degrees. Germane *et al.* (1989) and Miethke (1997) recorded torque errors of 10 and 5 degrees, respectively, with a vertical error of 1 mm.

A further essential factor in torque movement is the morphology of the dentition. Morrow (1978) found that it was subject to pronounced deviations, even in subjects with ideal occlusion. The angle between the long axis of the root and of the crown at an upper central incisor may vary widely (Carlsson and Rönnermann, 1973). In addition to bracket placement, the morphology of the dentition thus influences correct torque application (Germane *et al.*, 1989; Miethke, 1997).

The aims of the present study were to compare torque transmission from the archwire using polycarbonate brackets with and without metal slot inserts and to compare this with a metal bracket.

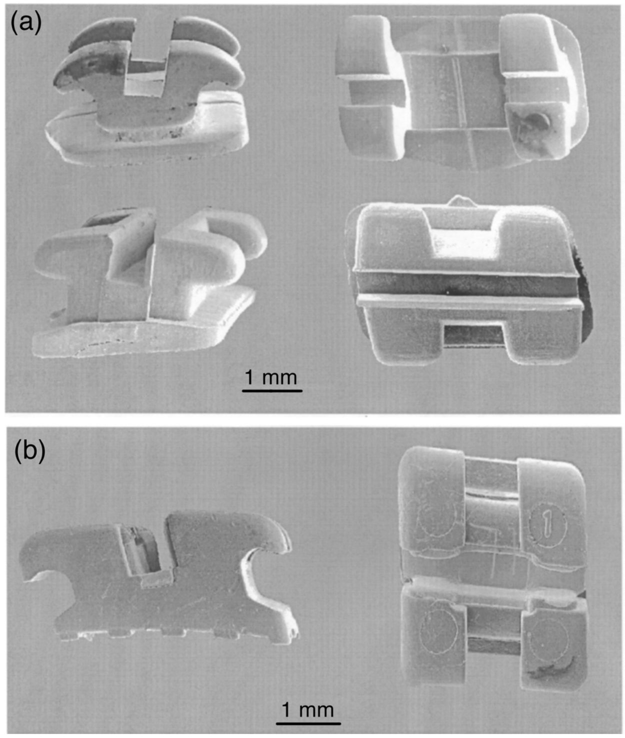

Figure 1 (a) Lateral and surface views of the Mini-Mono® metal bracket (top), the Brillant® polycarbonate bracket (bottom) and (b) the Elegance® polycarbonate bracket with a metal slot.

Materials and method

Three types of bracket were used in the trials. In addition to the Mini-Mono® metal bracket (Forestadent, Pforzheim, Germany), the Brillant® (Forestadent) polycarbonate bracket, improved specifically for torquing purposes and made of homogenous polyoxymethylene, was tested (Figure 1a). The second polycarbonate bracket, Elegance® (Dentaurum, Pforzheim, Germany) has a metal slot for improvement of torque transmission (Figure 1b). All brackets were tested in the 0.018 inch (0.46 mm) slot system. The smaller slot system was employed because of its more frequent use in daily practice. Measurement of the bracket slots from three polycarbonate brackets with and without a metal slot using a computer-compatible stereomicroscope with a scanner in the pilot phase of the investigation, revealed no slot width deviation from the 0.018 inch reference value (Gmyrek *et al.*, 2002).

Clinical simulation experiments

The clinical torque of an upper central incisor was simulated using the orthodontic measuring and simulation system (OMSS) developed at the University of Bonn. Torquing was performed on a levelled maxillary Frasaco model integrated into the OMSS. The OMSS is a measuring system developed specifically for investigating issues in the field of orthodontic biomechanics (Drescher *et al.*, 1991; Bourauel *et al.*, 1992). A schematic diagram and a detailed view are given in Figure 2(a) and (b). For simulation of tooth movement, the OMSS has a measuring table comprising a six-axis positioning table and a six-component force-torque sensor. This permits the complete force-torque vectors to be registered and the measuring points to be moved freely in space on optional path curves.

In the simulation experiments, the two polycarbonate brackets were compared with the Mini-Mono® metal bracket. From each type, five brackets were used. The stainless steel archwires used were 0.016×0.022 and 0.018×0.022 inch (Remanium®, Dentaurum). Each of the test brackets was connected to the torque-force sensor of the OMSS, positioned in place of tooth 21 (Figure 2b). A 20 degree buccal crown torque was then applied to the respective test bracket via the straight archwire which ran through all brackets of the dental arch and was ligated to them. The axes of tooth 21 are defined in Figure 3. The orthodontic movement of the incisor could then be calculated from the force system being applied at the test bracket, using a mathematical model of the OMSS (Bourauel *et al.*,

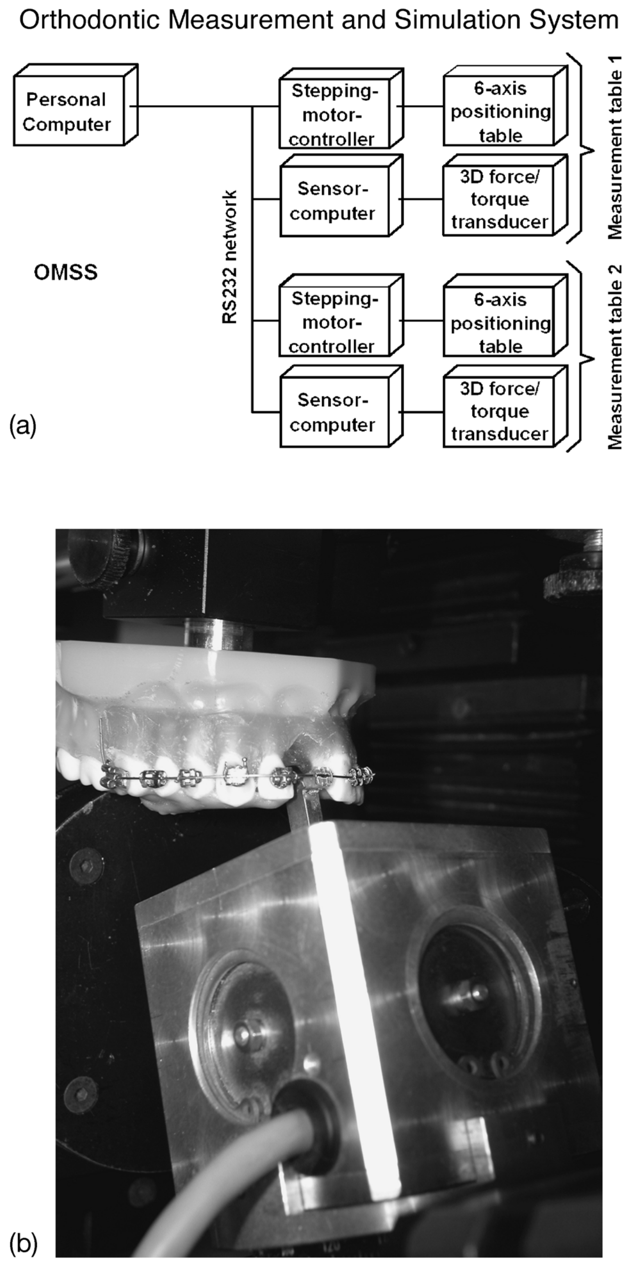

Figure 2 (a) The schematic structure of the orthodontic measuring and simulation system (OMSS) with its two measurement tables, each comprising a computer-controlled six-axis positioning table and a computer-controlled three-dimensional sensor (force-torque sensor). (b) A partial view of the OMSS with the Frasaco model at the six-axis positioning table and the test bracket at the force-torque sensor.

1992), and then executed by means of the stepping motor-driven positioning tables.

In general, the force systems comprise three forces and three moments. These six components are registered by the sensors of the OMSS. However, the reactive moments at the centre of resistance, resulting from the leverage effect of the force application point on the bracket, have to be added to the force systems acting purely on the bracket. These reactive moments are

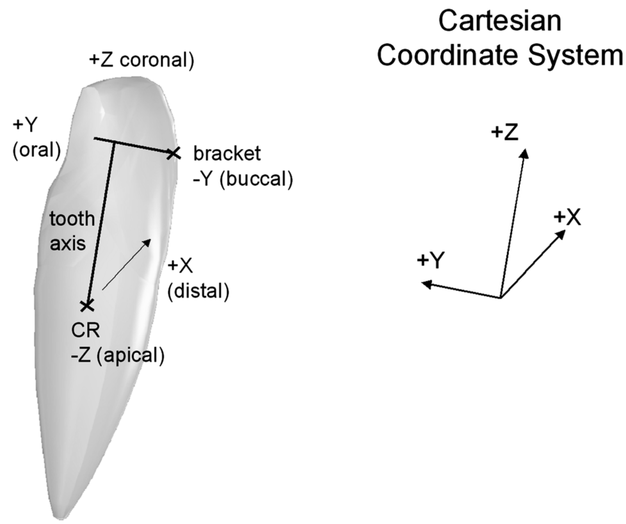

Figure 3 Definition of the tooth axes.

calculated automatically by the control programme of the OMSS and are thus entered into the simulated tooth movement. For this purpose the distance between the point of force application in the bracket slot and the centre of resistance of the tooth at the level of the first root third was set at 10 mm. By breaking the movement down into a large number of increments and repeating the 'measure force system–calculate movement–execute movement' cycle, even complex tooth movements as a reaction to applied force systems can be simulated. In the present investigation, incisor torquing was terminated when the torque fell below a value of 0.5 Nmm, which is the resolution limit of the torque-force transducer of the OMSS (Bourauel *et al.*, 1992). The movements were subdivided into a total of 700 increments, depending on the bracket. Each bracket–archwire combination was measured five times in total.

The data were processed using the WinSTAT for Excel program (version 2001.1, R. Fitch Software, Staufen, Germany). A Kolmogorov–Smirnov test for normality showed a non-normal distribution for all data sets except for the torque angle of the 0.018×0.022 Elegance® bracket, for which the moment fell below 0.5 Ncm. Consequently, all results are presented as box and whisker plots and statistical comparison of the torque situations was performed with the Mann–Whitney *U*-test (significance level $P < 0.05$).

Results

Plotting the measured torquing moments against the crown torque of the simulation measurements resulted in data similar to load/deflection curves of an orthodontic appliance. Figures 4 and 5 show examples of such curves for the three measured bracket types with the two wire dimensions. All relevant values, characterizing a given

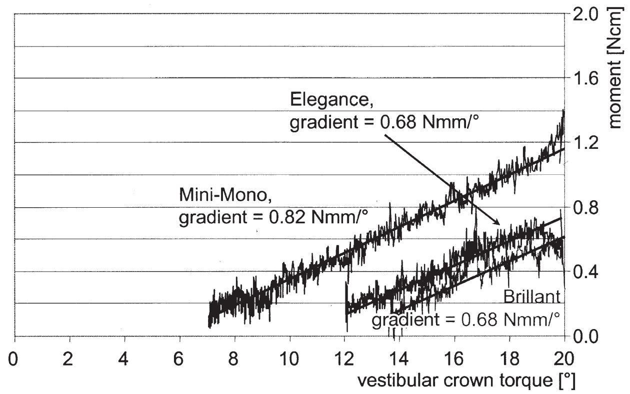

Figure 4 A graph showing the decline of moments dependent on the vestibular crown torque in Nmm/degree for the 0.016 × 0.022 inch archwire and the three bracket types.

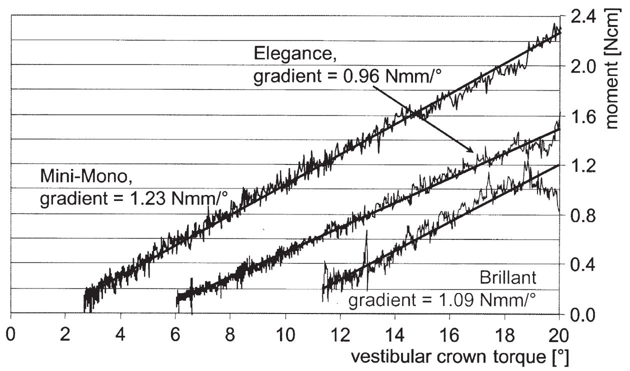

Figure 5 A graph showing the decline of moments dependent on the vestibular crown torque in Nmm/degree for the 0.018 × 0.022 inch archwire and the three bracket types.

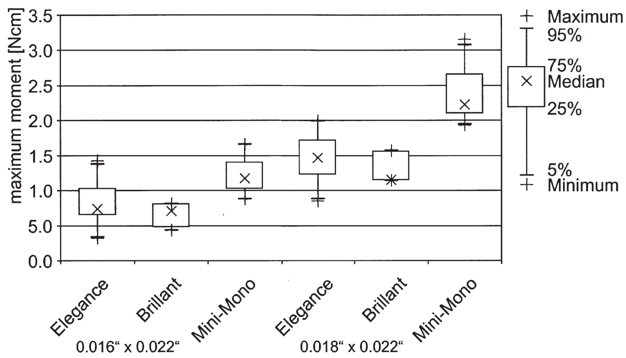

Figure 6 Maximum torquing moment in Ncm for the 0.016 × 0.022 and 0.018 × 0.022 inch archwires for the three bracket types.

Table 1 Simulation experiments. Median of maximum torquing moments (Mx) in Ncm depending on the bracket type and the archwire used and the results of Mann–Whitney *U*-tests for determining the difference between the maximum moments delivered by the brackets in combination with the respective archwire; significant at $P < 0.05$.

Bracket	0.016 × 0.022 inch	0.018 × 0.022 inch
Mini-Mono® (M)	1.17	2.22
Brillant® (B)	0.71	1.15
Elegance® (E)	0.74	1.47
$P < 0.05$	M/B: 0.000	M/B: 0.003
	M/E: 0.003	M/E: 0.000
	B/E: 0.029	B/E: 0.047

bracket–wire combination, can be taken from these plots. Some of these curves show an initial ascent of the moment curves to a maximum value, which means that the maximum torquing moment was not generated at the start of the experiment but after several degrees of vestibular crown movement. In these cases, the maximum moment was taken at the upper limit of the curves and the fit of the slope to determine the moment/torque ratio of the bracket–wire combination was taken at the linear part of the curve.

With 1.17 Ncm of torque for the 0.016 × 0.022 inch wire and 2.22 Ncm for the 0.018 × 0.022 inch wire, the median value of the maximum torquing moments recorded for the Mini-Mono® metal bracket was significantly higher than that for the Brillant® bracket; 0.71 and 1.15 Ncm, respectively (Figure 6, Table 1). The values for Elegance® with a metal slot were 0.74 and 1.47 Ncm, respectively, which was between the metal and polycarbonate brackets without a metal slot. However, the higher initial torques of the metal bracket were not achieved by either of the polycarbonate brackets. The results for both polycarbonate brackets were relatively close to one another and showed a marked variance, underlined by the higher significance values of the Mann–Whitney *U*-test for these combinations (Table 1) (test B/E = 0.029 and 0.047 for brackets in combination with the 0.016 × 0.022 and 0.018 × 0.022 inch wires, respectively). Nevertheless the maximum torque for all brackets differed significantly at the 0.05 level for both archwire dimensions.

The decline in torque and the torque course shown in Figures 4 and 5 were similar for the same wire dimensions, with a distinctly higher variance for the Elegance® bracket with the 0.016 × 0.022 inch wire. The slope of each individual moment/torque curve was calculated for each measurement of the six bracket–wire combinations and the results are presented in Figure 7. The median values of the moment/torque ratios ranged from 0.54 to 1.30 Nmm/degrees. It can be seen that the torque characteristics were determined primarily by wire dimension. This is underlined by the significance values of the Mann–Whitney *U*-test (Table 2): the differences between the Mini-Mono® and Elegance® brackets were not significant for the 0.016 × 0.022 inch archwire ($P = 0.087$). For the wire with the higher dimension, the difference between the Mini-Mono® and the Brillant® bracket was not significant ($P = 0.252$). However, with

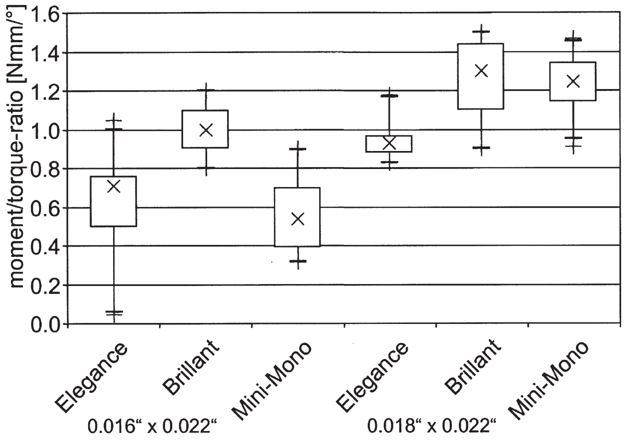

Figure 7 The moment/torque ratio in Nmm/degree for both wire types and the three bracket types.

Table 2 Moment/torque ratios (Nmm/degree) of the bracket–wire combinations investigated with the orthodontic measuring and simulation system and the results of the Mann–Whitney *U*-test for the different combinations.

Bracket	0.016 × 0.022 inch	0.018 × 0.022 inch
Mini-Mono® (M)	0.54	1.24
Brillant® (B)	1.00	1.30
Elegance® (E)	0.70	0.93
$P < 0.05$	M/B: 0.002	M/B: 0.252
	M/E: 0.087	M/E: 0.000
	B/E: 0.005	B/E: 0.003

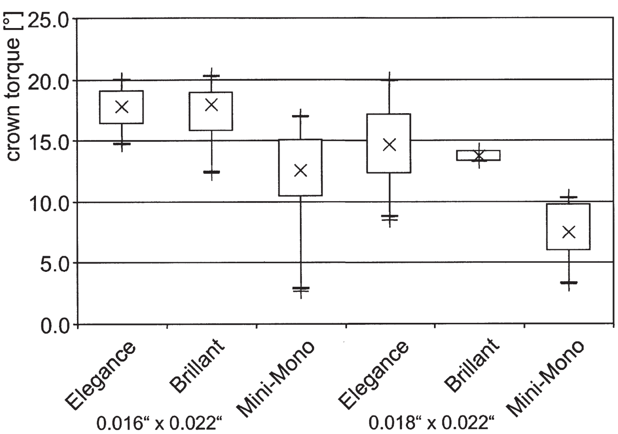

Figure 8 Crown torque in degrees at which the torquing moment fell below the minimum value of 0.5 Ncm.

the Brillant® bracket the moment/torque ratio was slightly higher in each case, i.e. the moment declined more sharply with a decreasing torque angle than with the Mini-Mono® and Elegance® brackets.

Figure 8 displays the torque angles at which the torquing moment fell below the minimum limit of 0.5 Ncm. With the 0.018 × 0.022 inch archwire, the metal

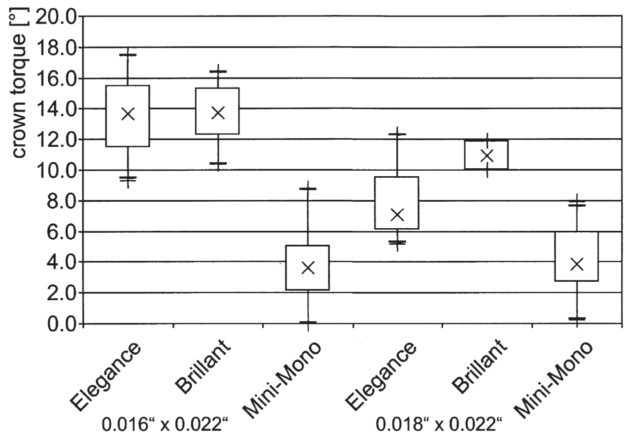

Figure 9 Maximum correction of crown torque (degrees).

bracket reached this point at a median torque angle of 7.5 degrees, while the polycarbonate brackets delivered this minimum torque at angles of 13.7 and 14.7 degrees, respectively. With the 0.016 × 0.022 inch archwire this limit was reached at 12.6 degrees for the metal bracket, whereas the moment delivered by the polycarbonate brackets fell below 0.5 Ncm at angles of 17.8 and 17.9 degrees.

Continuing the simulation until the torquing moment reached a value as low as 0.5 Nmm, i.e. there was no more measurable torquing effect at the bracket, the final torque angles of the brackets in combination with the 0.018 × 0.022 inch wire were as follows: 3.8 degrees for the metal bracket, 10.9 degrees for the Brillant® bracket and 7.1 degrees for the Elegance® bracket with a metal slot (see Figure 9). Using the 0.016 × 0.022 inch archwire, the values were: 3.6 (Mini-Mono®), 13.7 (Brillant®) and 13.7 degrees (Elegance®) (Figure 9).

Discussion

Polycarbonate brackets, due to their viscoelastic characteristics, are reputed not to torque teeth adequately. Dobrin *et al.* (1975) reported marked deformation and low torque values with polycarbonate brackets. This observation was confirmed by Feldner *et al.* (1994) who reported that filler-reinforced polycarbonate brackets had clinically more acceptable torque values than conventional polycarbonate brackets due to their markedly lower deformation. In the above studies, activation measurements were also performed on individual brackets.

This lower deformation was partly confirmed by the results of the activating experiments in the present investigation. However, the deformation of the polycarbonate brackets descended, especially for Brillant®, with a decreasing torque angle. In total the measured behaviour of the moment/torque ratios of the different bracket–wire combinations was highly inconsistent.

However, a more detailed and, in particular, a closer statistical analysis of these characteristics would involve measuring a large number of bracket–wire combinations, to identify the different influences. This was beyond the framework of the present study. In a former investigation with the same type of polycarbonate brackets, measurements were taken at the slots after the torquing trials revealed no permanent deformation. The pre- and post-torque slot widths of the polycarbonate brackets differed only insignificantly and are thus of no clinical relevance (Gmyrek *et al.*, 2002). One striking feature, however, was that the slot widths showed convergence upwards and were thus slightly below the reference value of 0.018 inch (0.46 mm) at the upper edge. This was confirmed by the manufacturer, Forestadent, who attributed it to the tension within the bracket resulting from the injection-moulding procedure (Gmyrek *et al.*, 2002).

The higher torque loss recorded for the two polycarbonate brackets with both archwires must have been the outcome of elastic deformation of the slot. The metal slot did not have the anticipated effect in the polycarbonate bracket. Additionally, this elastic bending complies with data issued by the manufacturer on the 'resilience' of this bracket. This is seen as positive with respect to decreasing force and a consequent reduced risk of root resorption. On the other hand, the corresponding restoring or 'shock absorber' effect claimed by the manufacturer is not to be expected, so that any force discharge from the polycarbonate is unlikely. This means that the transmitted torquing moment can be less reliably predicted than for metal brackets. A strength deficit is, however, revealed by the higher torque losses of the polycarbonate brackets, slightly less with a metal slot, compared with the metal bracket.

For simulation of a clinical torque situation with the OMSS, tooth 21 was replaced by a sensor to which the test bracket was attached. A 20 degree buccal crown torque was then applied to the test bracket by means of a straightwire running through all brackets of the dental arch. After insertion into the bracket slot, this led to immediate twisting of the wire and, in the case of the polycarbonate brackets, to slot expansion visible with the naked eye. As is customary in the clinical situation, the ligature was then attached to the bracket to secure the continuous archwire.

In previous activation measurements performed on a precision lathe (Gmyrek *et al.*, 2002), the archwire was ligated tightly in the bracket prior to actual torquing. For simulation of the clinical situation, however, a substantial proportion of the force had to be lost in the immediate bending of the polycarbonate bracket. This may account for the significantly lower torquing moments recorded for the polycarbonate brackets with and without a metal slot in comparison with the activating experiments. Although subsequent measuring of the slot widths with a stereomicroscope revealed no increase in size, the metal bracket also had significantly lower torquing moments in the simulation experiments than in the activating experiments. With all types of bracket, this was due to the greater play of the twisted archwire in the simulated clinical situation. On average, the interbracket distance was 6 mm from the clamped test bracket to neighbouring teeth. In the clinical dental arch model in the OMSS the neighbouring brackets permitted additional play. The actual torque loss was thus well above the values registered in the previous *in vitro* activating torque experiments, as it occurred not only at the teeth to be moved but also at the anchorage teeth.

Another reason for the low torques and their rapid decline with a decreasing torque angle was that the torque was generated under simulated clinical conditions. In the twisting process, the torque is generated by the outside edges of the wire. This, however, also results in a change in length in terms of a shortening of the archwire, so that a continuous archwire is deformed. This leads to auxiliary forces which on one hand generate a counter-torque in the anterior segment and at the incisor by means of the leverage effect, and on the other act in the other two dimensions in the form of torques and are thus lost for the torquing moment. The simulated tooth then starts the torque movement and reacts very rapidly to the forces. It 'follows' them, and the forces as well as the torques disappear. An exact mathematical analysis is no longer feasible on account of this mechanical sequence. The pronounced complexity of this situation is reflected in the torque curves of the two polycarbonate brackets (Figures 4 and 5). Based on the adjusted buccal crown torque of 20 degrees, the torquing moment first increases with a decreasing torque angle and then declines in a manner similar to the steel bracket. This can only be due to the polycarbonate bracket and archwire being deformed in the course of clamping. The outcome is that the archwire cannot transmit the complete torquing moment to the bracket wing on account of the initial force system. It is only in the course of the simulated movement, in which the incisor follows the force system as described above, that the deformation diminishes as the auxiliary forces are rapidly reduced. To the same extent, the deformation of the archwire is reduced, approaching the ideal twisting required for the generation of a torque.

Although the OMSS model comes very close to the clinical situation, it fails to take into account some factors which have an additional influence in practice: the long-term effect of the torque on the brackets, the influence of saliva on the bracket material and the side-effects on the adjacent teeth. These had to be disregarded in the present investigation because the teeth were anchored relatively rigid in the Frasaco model.

The complex mechanical situation of the incisor torque and the factors influencing it illustrate that *in vitro* trials are no substitute for clinical testing.

Conclusions

The results of the present study suggest that both tested polycarbonate brackets and the Mini-Mono® metal bracket can be recommended for torquing purposes. However, no conclusion can be reached regarding their long-term strength in clinical application.

Against the background of the high torque losses of the polycarbonate brackets in the activation experiments, however, the torque values programmed in the straightwire technique appear to be questionable, as they remained virtually ineffective on a scale of 13 degrees with the polycarbonate brackets. Data should be provided by the manufacturer on the predicted bending to be compensated for by additional torque or torque values should be omitted from the technical data.

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