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A model for evaluating friction during orthodontic tooth movement

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SUMMARY Orthodontic forces for sliding tooth movement during space closure are applied at a distance from the centre of resistance of the teeth. For this reason, the teeth will tip until contacts are established between the archwire and diagonally opposite corners of the bracket wings. They will also rotate until the wire contacts opposite corners of the ligature tie or the buccal shield with self-ligating brackets, and the base of the slot. Frictional forces measured with models that do not enable such movements may therefore not be representative of the clinical condition. To test this hypothesis, a dentoalveolar model that allowed accurate reproduction of the width of a material of similar elastic properties as the periodontal ligament (PDL) was fabricated. In addition, a device was designed that allowed accurate adjustment of the bracket slot in all three planes of space during mounting of the model in an Instron machine. Frictional forces during sliding of ceramic brackets with 0.022 × 0.028-inch bracket slots along 0.019 × 0.025-inch stainless steel wires were tested using models with simulated PDL widths of 0.00, 0.33, 0.67, and 1.00 mm.

ANOVA detected a significant effect of PDL width on mean frictional force ($P < 0.001$). Pairwise comparisons at 0.05 significance level indicated no differences between the models without PDL and those with a width of 0.33 mm, and between models with PDL widths of 0.67 and 1.00 mm. However, the two models with smaller widths produced significantly lower frictional forces.

Introduction

Friction is the resistance to motion when one object moves tangentially against another (Besancon, 1985). The normal force is the perpendicular component of the force acting on the contacting surfaces (Giancoli, 1980). The coefficient of friction for a given material surface is a constant, which may be dependent on the roughness, texture, or hardness of the surfaces (Besancon, 1985). The actual frictional force is the product of the coefficient of friction and the normal force. In order for one object to slide against the other, the force application must overcome the frictional force.

Orthodontic forces are applied at a distance from the centre of resistance of the teeth. For that reason, the teeth will tip and rotate in the direction of force application. With fixed appliance therapy, the tipping will be counteracted once contacts are established between the archwire and diagonally opposite corners of the bracket wings (Figure 1A,C). Similarly, the rotation will be counteracted once contacts are established between the archwire and diagonally opposite corners of the ligature tie, or the buccal bracket shield with self-ligating brackets and the base of the slot (Figure 1B). These movements will occur upon force application. Provided the orthodontic forces overcome the frictional forces produced at the points of contact between archwire, bracket slot, and ligature, the teeth will slide along the archwire. If the archwire does not deform or the ligatures do not break, further tipping and rotation will not occur during sliding tooth movements.

The ongoing development in material and design of orthodontic brackets and archwires is likely to continue. Due to differences in bracket slot configuration and in surface texture of

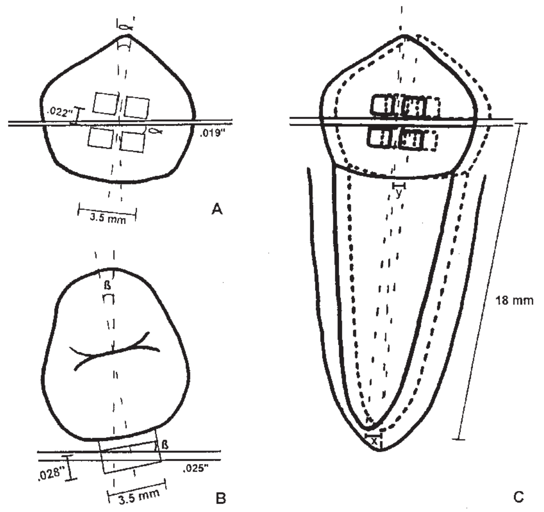

Figure 1 Initial tooth displacement relative to a 0.019 × 0.025-inch archwire upon application of forces to an orthodontic bracket with a slot size of 0.022 × 0.028-inch and a mesiodistal width of 3.5 mm (not drawn to scale). (A) Amount of tipping (α), 1.25 degrees. (B) Amount of rotation (β), 1.25 degrees. (C) Maximum displacement of the root of an average size premolar, 0.4 mm (×) with the centre of rotation at the bracket, and 0.3 mm (y) with the centre of rotation at the root apex.

the materials, different bracket and archwire combinations may produce different amounts of friction. Standardized clinical conditions for accurate evaluation of friction are difficult to establish. For this reason, the vast majority of the information is derived from laboratory experiments. However, such experiments should utilize a model that reproduces the clinical condition in order to generate valid inferences.

The most frequently used experimental model for evaluation of orthodontic friction consists of simulated teeth mounted in a fixed medium (Andreasen and Quevedo, 1970; Frank and Nikolai, 1980; Peterson *et al*., 1982; Baker *et al*., 1987; Kusy and Whitley, 1990; Pratten *et al*., 1990; Kusy *et al*., 1991; Prososki *et al*., 1991; Keith *et al*., 1993; Sims *et al*., 1993, 1994; Dickson *et al*., 1994; Downing *et al*., 1994; Saunders and Kusy, 1994; Tselepis *et al*., 1994; DeFranco *et al*., 1995). However, such models do not permit the initial tipping and rotation that occurs clinically. Therefore, the amount of friction measured when archwires are pulled through bracket slots bonded to the mounted teeth may reflect variations in normal force due to differences in tension from the ligature ties, rather than differences in frictional coefficients. Few studies have utilized models more representative of the clinical condition. Tanne *et al.* (1991) applied constant forces to teeth invested in wax, and interpreted an increase in tooth displacement as a reduction in friction. One potential difficulty with such models is that wax may not maintain constant resistive properties throughout the material, due to temperature variations. Angolkar *et al.* (1990), Kapila *et al.* (1990), Bednar *et al.* (1991), Tanne *et al.* (1994), and Vaughan *et al.* (1995) utilized a model that permitted the initial tipping movement. However, the rotational movements appeared not to be reproduced. The models may therefore not be valid for testing of differences in friction between self-ligating brackets and brackets that need ligation. Drescher *et al.* (1989) cemented brackets to a brass bar imbedded in an elastic rubber foam, allowing three-dimensional (3D) initial movement upon force application. However, the model was designed to produce archwire deformation, which may introduce side-effects such as binding and pinching. Such deformation may not occur during space closure under ideal conditions.

The purpose of this study was to design a laboratory model for standardized measurement of frictional forces during sliding mechanics under simulated clinical space closure, and to test the hypothesis that the frictional forces with such a model are larger than with a model that does not allow initial tipping and rotation.

Materials and methods

Dentoalveolar model

An indirect technique was used to reproduce models with identical width of a material of similar elastic properties as the periodontal ligament (PDL) between the root of an ivorine maxillary second premolar (Columbia Dentoform, Long Island, NY, USA) and an acrylic socket. Initially, lead foil of thickness 0.33, 0.67, or 1.00 mm (two, four, or six layers, respectively, taken from occlusal radiograph film) was tightly adapted to the entire root surface of the ivorine tooth (Figure 2A). Following light lubrication with petroleum jelly (Topco Association, Skokie, IL, USA), the foil covered root was inserted into a wet mix of acrylic (Jet, Lang Dental, Wheeling, IL, USA) and poured into a right circular Plexiglas cylinder. Excess acrylic was trimmed upon curing, and a heavy body viscosity polyvinylsiloxane impression material (Express STD, 3M, St. Paul, MN, USA) was used to index the cylinder and the occlusal, lingual, mesial, and distal surfaces of the tooth crown (Figure 2B). The index and the foil covered tooth were then separated from the acrylic socket. A 'V'-shaped notch was carved in the Plexiglas and the index was relined with a light body viscosity polyvinylsiloxane impression material (Extrude, Kerr, Romulus, MI, USA). Following removal of the foil (Figure 2C) and injection of a light body polyvinylsiloxane impression material (Extrude) into the acrylic socket, the tooth, and the relined index were refitted to the Plexiglas cylinder. Excess material was trimmed upon setting.

Test apparatus

A device was designed that allowed adjustment of the insertion level in the sagittal plane and rotation in the frontal plane of the Plexiglas cylinder during mounting. In addition, the cylinder bearing part of the device could be rotated in the sagittal plane (Figure 3A). The base of the device allowed attachment to a universal testing machine (Instron Corp., Canton, MA, USA) through a vice, adjusting the level of insertion in the frontal plane (Figure 3A). If necessary, the vice could be adjusted in the sagittal plane. The device therefore ensured passive connection between archwire, and Instron machine in the sagittal and frontal planes.

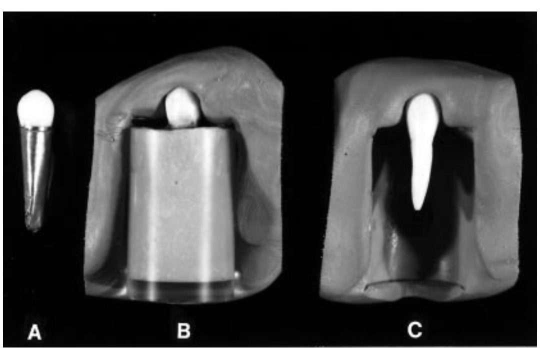

Figure 2 Construction of dentoalveolar model. (A) Ivorine maxillary second premolar with lead foil of a standardized thickness adapted to the entire root surface. (B) Foil covered root inserted into acrylic contained in right circular Plexiglas cylinder. Cylinder and crown indexed in polyvinylsiloxane. (C) Tooth and index upon separation from cylinder and removal of foil.

Test procedure

Initially, only four dentoalveolar models were fabricated, three with simulated PDL spaces (0.33 mm, 0.67 mm, and 1.00 mm) and one without simulated PDL space by inserting a tooth without lead foil into the wet acrylic poured into the Plexiglas cylinder. A ceramic bracket with a 0.022 × 0.028-inch slot (Transcend, 3M-Unitek, Monrovia, CA, USA) was bonded to the mid-buccal surface of each tooth using cyanoacrylate cement (Krazy glue, Borden, Inc., Columbus, OH, USA). One unit of a power chain (short length, 3M-Unitek) was placed over each bracket before a 45-mm straight length of a 0.019 × 0.025-inch stainless steel archwire (Hi-T II, 3M-Unitek) was fitted to the bracket slot and ligated passively to the tie wings with a 0.010-inch pretwisted stainless steel ligature (Quik-Tie, Masel, Bristol, PA, USA). A pilot session was then conducted, mounting each of the four models to the test apparatus and the Instron machine in random order. A passive connection between archwire, and load cell in the sagittal and frontal planes was obtained each time (Figure 3A,B). The archwire was clamped to the load cell via a ball bearing connection piece, allowing passive connection in the horizontal plane (Figure 3B). Following this procedure, the adjacent unit of the power chain was attached to a hook 8 mm from the bracket (Figure 3A) and a drop of artificial saliva (Saliva Substitute, Roxanne Laboratories, Columbus, OH, USA) was placed on the bracket. The tests were run at a speed of 0.5 mm/minute for 4 minutes, producing 2 mm of archwire movement. The data were recorded on an XY recorder. The X-axis

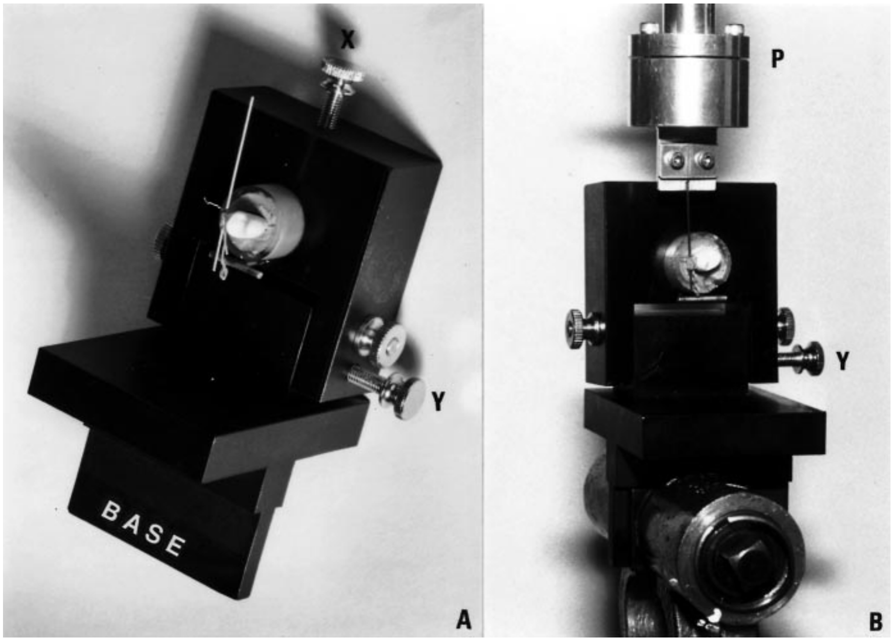

Figure 3 Experimental model. (A) Test apparatus. Screw (X) adjusted level of insertion in plane A and rotation in plane B. Screw (Y) allowed rotation in plane A. Note attachment of power chain to bracket base and metal hook. (B) Mounting of test apparatus in Instron machine. Base of test apparatus allowed adjustment in plane B. Ball bearing assembly (P) allowed passive connection in plane C.

recorded wire movement in millimetres, and the Y-axis the frictional force between the bracket and the archwire in grams.

After completion of the four tests of the pilot session, three new models were made of each type. These 12 models were tested in random order as described. Similarly, two additional test sessions were conducted with 12 comparable models. Prior to each additional session, all light body silicone material was removed from each acrylic socket of the dentoalveolar models with simulated PDL spaces. The crowns of new ivorine premolars were then fitted to each index. Finally, each index was fitted to each respective Plexiglas cylinder following injection of new light body material. Bracket bonding, wire ligation, and mounting in the Instron machine were undertaken as before. For the models without simulated PDL space, the old brackets were removed from the teeth mounted directly in the acrylic and new brackets were bonded prior to each new session.

Data analysis

The XY recordings were coded. One frictional force was calculated for each of the four pilot and each of the 36 regular tests by averaging 16 recordings, 12 seconds apart on the Y-axis, starting at 1 minute. Descriptive statistics were calculated for each PDL width. A mixed effect analysis of variance (ANOVA) model was fitted with a fixed width effect, a random session effect, and a random width by session interaction term. Also, pairwise comparisons at an overall 0.05 significance level were made between the PDL widths using Tukey's method under the full mixed effects model. Initially, the four pilot tests of the first session were omitted so that the resulting balanced design could be analysed using ANOVA. Then a more complicated approach was used to account for the unbalanced design when including the pilot session (Satterthwaite, 1946). This approach adjusts the sums of squares and degrees of freedom used in the *F*-tests associated with the ANOVA.

Measurement error

The reproducibility of the calculation of frictional force from the XY recordings was assessed by statistically analyzing the difference between double calculations made at least 1 week apart on 20 recordings. The error was calculated from the following equation:

$$s_x = \sqrt{(\Sigma D^2/2N)}$$

where D is the difference between duplicated calculations and N is the number of double calculations (Dahlberg, 1940). The error was 0.19 g, which represented 0.27 per cent of the mean friction value of 71.6 g, over all sets of bracket-archwire combinations.

Results

The calculated frictional force for each dento-alveolar model varied among the tests at each session, as well as from session to session (Table 1). The ANOVA model did not detect a significant effect of the session or the width by session interaction term (Table 2). However, the effect of PDL width on mean frictional forces was significant (P = 0.009, Table 2). Adjusting for multiple comparisons at an overall significance of 0.05, Tukey's method of pairwise comparisons of mean frictional forces at each session suggested significantly different PDL widths if the difference in frictional force was greater than 35.23 g. Specifically, the pairwise comparisons indicated no significant differences between the models without PDL and those with a width of 0.33 mm, and between models with PDL widths of 0.67 and 1.00 mm. However, the two models with smaller widths were significantly different from the two with larger widths. The results from the data analysis that accounted for the unbalanced design when including the results from the pilot session, agreed with those obtained from the ANOVA analysis. In particular, the F-test for the width effect was significant (F = 10.66 with 2.94 and 9 degrees of freedom, P = 0.003).

Table 1 Mean (SE) of calculated frictional forces (g) per test session using models with four simulated widths of the periodontal ligament (PDL).

PDL width (mm)	Session 1 4 Tests	Session 2 12 Tests	Session 3 12 Tests	Session 4 12 Tests	All sessions
0.00	10.31 (NA)	7.04 (3.12)	8.83 (6.31)	10.15 (5.29)	8.84 (2.66)
0.33	21.25 (NA)	20.00 (5.36)	10.81 (0.59)	14.02 (2.37)	15.58 (2.22)
0.67	30.19 (NA)	53.02 (13.41)	76.69 (10.64)	29.60 (8.82)	50.81 (8.67)
1.00	97.69 (NA)	45.92 (15.84)	54.12 (13.47)	51.06 (3.90)	55.10 (7.96)
All	39.86 (19.70)	31.49 (7.76)	37.61 (9.81)	26.51 (5.59)	

NA, not applicable.

Table 2 ANOVA model summary.

Effect	Degrees of freedom	Sum of squares	F value	P value
Width	3	14558.19	10.43	0.009
Session	2	782.00	0.84	0.477
Width × Session	6	2791.55	1.50	0.221
Error	24	7443.15		
Total	35	25574.89		

Discussion

The hypothesis that different experimental models generate different frictional forces during simulated tooth movement with identical bracket and archwire combinations was confirmed. Sliding tooth movements clinically are always preceded by tipping and rotation of the teeth. Provided levelling and alignment are completed, and archwire deformation does not occur, the amount of tipping is determined by the discrepancy in vertical dimension between archwire and bracket slot. Similarly, the amount of rotation is determined by the discrepancy in horizontal dimension between archwire and bracket slot. The latter statement is based on the assumption that the ligature ties are flush with the profile of the bracket wings. The results strongly suggest that studies on differences in friction among different bracket and archwire combinations do not allow inferences to the clinical situation, unless the models that are used simulate the initial tipping and rotation movements.

Using 0.022 × 0.028-inch bracket slots with a mesiodistal width of 3.5 mm and 0.019 × 0.025-inch archwires, the maximum narrowing of the PDL will be approximately 0.4 mm during the initial tipping and rotation of a premolar of average length (Figure 1C). The average PDL space in humans is about 0.2 mm and teeth in function tend to have a wider space, particularly in the cervical and apical portions (Coolidge, 1937). However, the distance between the root surface and the alveolar socket may double or triple during periods of orthodontic tooth movement (Svanberg, 1974). In the present study, 0.67 mm was chosen as an experimental PDL width to simulate the clinical situation with contact between archwire, bracket slot, and ligature in virtually every case on application of orthodontic forces for space closure. The wider width was included to test the hypothesis that an increase in PDL space beyond what is needed to allow such contacts is of minor significance for the measurements of friction. The narrower spaces were included to test the hypothesis that the friction measurements will be reduced if the PDL space is too narrow to allow contacts between archwire, bracket slot, and ligature. The results of this study suggest that the majority of the dentoalveolar models with simulated PDL widths of 0.67 and 1.00 mm may have permitted contacts between archwire and diagonally opposite corners of the bracket slots, and between archwire, bracket base, and ligature during the tests, explaining the comparable frictional forces with those models. Similarly, the dentoalveolar models with a simulated PDL width of 0.33 mm may not have allowed maximum contact between archwire, bracket slot, and ligature during the course of the experiment. This may explain why the statistical analysis in this study failed to detect any difference in frictional force between these models and those without a simulated PDL. However, with the centre of rotation midway between the cemento-enamel junction and root apex, the narrowing of the PDL may be as minimal as 0.2 mm during the initial tipping. Accordingly, contacts may have been established between archwire and slot with some of the models with a simulated PDL width of 0.33 mm, explaining the increased range of frictional force measurements with these models (Table 1).

The design of the test apparatus in this investigation allowed adjustment of the bracket slots in all three planes of space during mounting of the dentoalveolar models in the Instron machine (Figure 3A,B). In this experiment the aim was to simulate sliding tooth movement following complete levelling and alignment, with passive fit of the archwire in the bracket slot prior to orthodontic force application. However, by changing the inclination of the bracket slot relative to the direction of movement of the test apparatus, this model may also be used to evaluate the increase in frictional force with standardized amounts of active tip and rotation concomitant with sliding tooth movement. In order to evaluate the increase in frictional force with standardized amounts of torque, the ball bearing in the connection piece to the load cell would have to be removed.

Despite efforts to standardize the testing conditions and to minimize method error due to variation in angle between bracket slot and direction of movement (Figure 3A,B), the variation in frictional force measurements with similar dentoalveolar models was considerable (Table 1). For example, the models with a simulated PDL width of 0.67 mm showed frictional forces ranging from 13.7 to 89.5 g. One likely explanation may be bias due to differences in normal force caused by variations in pressure from the ligature ties. Several studies have documented that the increase in normal force due to tight ligation will cause an increase in the frictional force (DeFranco *et al*., 1995; Frank and Nikolai, 1980; Keith *et al*., 1993; Sims *et al*., 1993; Stannard *et al*., 1986). To reduce the potential for such bias, all ligation was carried out by one author (BL) according to a standardized procedure aiming to avoid increase in the normal force. For the same reason, the power chain was applied under the wire, around the bracket base (Figure 3A) to avoid an increase in the normal force when applying the simulated orthodontic force. These precautions suggest that the large variation may be due to random errors inherent in studies of this type. One source may be manufacturing errors, causing variation in surface quality and size of archwires and bracket slots. Another possibility may be inconsistencies in the measurements by the load cell of the Instron. To avoid such error, however, the Instron machine was calibrated with a 100-g load before each test session.

The hook that was placed 8 mm from the bracket allowed standardization of the active forces to the tooth (Figure 3). This distance was chosen to simulate active closure of a premolar extraction site at a stage shortly after levelling and alignment, using a power chain where adjacent holes are attached to brackets bonded to each tooth in the arch. This model therefore reproduced the forces commonly used under normal clinical conditions. Variation in force level as a result of differences in elastic properties among the power chain units or of inconsistencies in mesiodostal placement of the brackets was considered to be minimal.

Because of the variation among the tested samples, testing of the models in random order, and with identical proportion of the models in each test session may be considered important. It may be criticized that inclusion of the first pilot session created an unbalanced design because the number of models was reduced. However, the results from the analysis that adjusted for the unbalanced design caused by including the pilot session agreed with those obtained from the ANOVA analysis of the three test sessions.

A further criticism may be that the material used to simulate the PDL may have physical properties different from actual tissues (Yousefian *et al*., 1996). On the other hand, the material did allow tipping until contact was established between bracket and archwire when models with PDL spaces of 0.67 and 1.00 mm were used. The assumption that this occurs clinically may lead to the speculation that the actual PDL properties are of minor significance for the development of frictional forces. Another criticism may be that no efforts were made to simulate occlusal forces with the laboratory model. However, the clinical effects of saliva were duplicated and, for standardization purposes, artificial as opposed to human saliva was used.

Conclusions

Laboratory studies of frictional forces during sliding tooth movement should utilize a model that simulates the initial tipping and rotation that occurs during clinical conditions until contact is established between bracket slot, ligature, and archwire. In addition, a device should be manufactured that allows adjustment of the bracket slot in all three planes of space during mounting in the testing machine.

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