Orthodontic forces generated by a simulated archwire appliance evaluated by the Finite Element Method

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One of the essential concerns of orthodontic tooth movement is the magnitude and direction of orthodontic forces exerted on abutment teeth. Overstressing dental abutments can result in osseous cavitation, root resorption, and residual mobility, while low force applications often can be responsible for prolonged treatment time, predisposing patients to dental caries or periodontal disease. Therefore, knowledge of the magnitude of forces generated on abutment teeth during orthodontic treatment can be useful to practitioners. A number of dental investigations have been carried out utilizing finite element theory examining implant-bone interactions,[1] ceramic-metal bonds,[2] thermal expansion in dental restorations,[3,4] and tooth loading.[5,6] Several preliminary studies have focused on orthodontic applications including tooth movement,[7-9] and the relationships of orthodontic movement to supporting tissues.[10-11] The purpose of this research was to study one way in which the finite element method (FEM)[12] might be applied to an orthodontic problem to evaluate initial forces generated by an orthodontic appliance on a simulated dental malocclusion.

Materials and methods

The FEM was employed to determine the intensity and direction of forces stress distribution generated on teeth by the initial placement of a preangulated, pretorqued, (bracket-type) orthodontic appliance (Unitek Twin Torque, Monrovia, CA) utilizing rectangular titanium-nickel alloy archwire (Unitek Nitinol, Monrovia, CA; Elastic modulus = 4 x 106 psi) of thicknesses ranging from 0.016 x 0.025 to 0.024 x 0.025 in. The FEM analysis was based on a direct stiffness method formulation, which allowed for efficient treatment of the highly statically indeterminate configuration. The wire was modeled as a space-frame composed of beam elements with 6 degrees of freedom per node. Figure 1 portrays a representative beam element and the associated positive nodal forces and displacements with respect to the local Cartesian

Abstract

The finite element method has been used to determine the stress distribution generated by the initial placement of a simulated preset bracket-type orthodontic appliance utilizing titanium-nickel alloy archwire.

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Key Words

Finite element method • Orthodontic forces • Appliance analysis • Stress analysis • Direction of forces

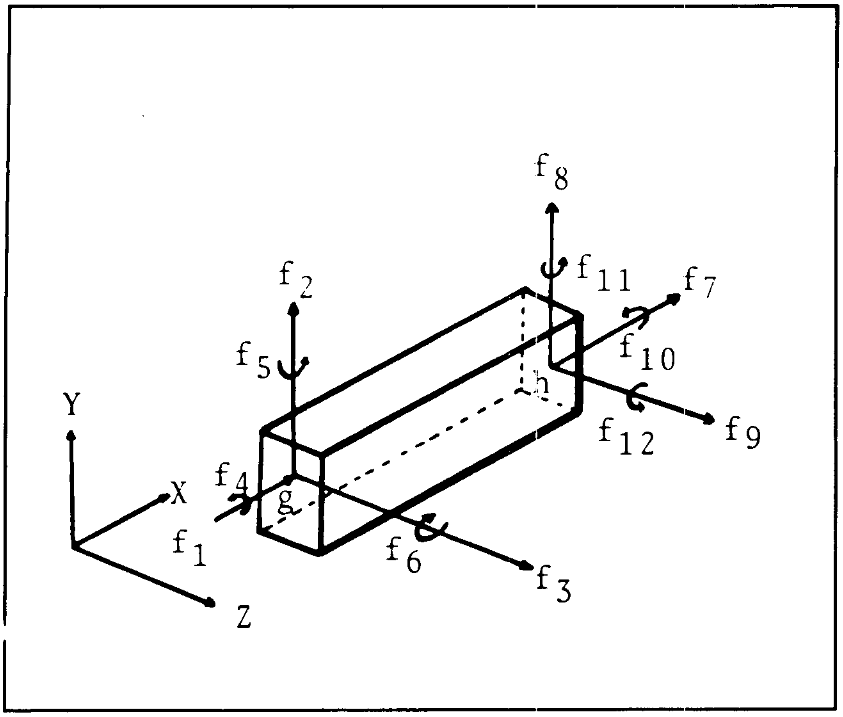

Figure 1

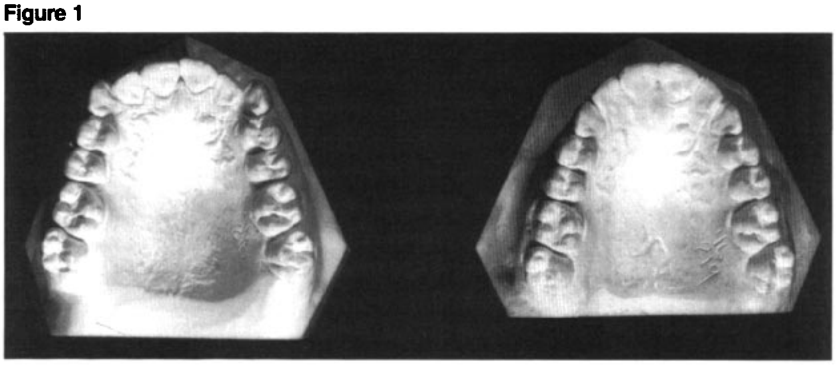

Figure 2

Figure 1
Space-frame beam elements with nodal degrees of freedom.

Figure 2
Pre- and post-treatment maxillary casts used for deriving abutment locations.

system (X,Y,Z). The bracket wire contact was simulated as a three-dimensional connection allowed to freely displace along the plane tangent to the bracket-wire line of contact. The element stiffness was developed under the assumption of linear theory and small deformations. In order to minimize the computational effort and input data maintaining finite element method modeling requirements, one beam element between two adjacent abutments was used to model the alloy wire.

A maxillary pre-treatment gypsum model of a malocclusion case was selected for this study and was mounted on a Hanau articulator. The three-dimensional coordinates of all abutment teeth bracket positions were located. After correction, the post-treatment abutment locations were identified from a model and recorded (Figure 2). The change in spatial location for each abutment tooth was derived by comparing respective global positions of bracket locations on pre-treatment and post-treatment casts. This data along with the wire physical properties, and bracket settings, provided the input for the FEM procedure on an IBM 3081-GX Mainframe computer.

The appliance selected to simulate this orthodontic treatment model included preset bracket torques, angulations, and slopes as available from the manufacturer (Table 1).

Results

The analysis revealed initial abutment tooth vector forces ranging from 0.21 to 6.92 Newtons (1 Newton = 102 grams/forces). The direction of the net forces with reference to the global Cartesian system can be evaluated from the direction cosines given by $\cos(F_I,x_I)=/(F_1^2+F_2^2+F_3^2)1/2$ where Fi (i=1,2,3) denote the components of the net forces with respect to the x1, x2, and x3 axes respectively. At the joints of each beam element, the local magnitude and direction can readily be evaluated from the components of the net forces through standard finite element procedures. The range of each directional force produced by archwire applications of various dimensions on abutment teeth is shown in Table 2. Displacements included mesio-distal, facial-lingual, intrusion-extrusion, two-dimensional tilting, and axis rotation.

An examination of the net forces produced on each abutment tooth are shown in Table 3. As the thickness of the wire increased, the net forces on the abutment teeth correspondingly increased. However, the net direction of forces exerted by this appliance on the dental arch were not significantly affected by different wire dimensions within the treatment load range examined in this case.

Discussion

The case examined in this project exhibited a Class I molar relationship with maxillary crowding reflecting an arch length deficiency. The appliance selected for this simulated treatment plan has been previously described. Given the initial coordinates of the abutment teeth and the proposed abutment positions, it was possible to calculate the forces generated on abutment teeth from the initial appliance placement.

An optimum load range for movement of a tooth varies with several considerations such as the number of roots, gross size, the movement distance and direction, and other frictional requirements. However, a reasonable range might be from 20 to 300 grams[13] beyond which damage to the supporting tissues or root resorption may become a problem. Vector facial-lingual tilting forces were generally of low magnitude, perhaps due to the fact that the appliance bracket presetting was 0°all cases. However, since sev-

TABLE 1

Tooth	Pre-Treatment Coordinates X	Y	Z	Appliance Torque	Pre-set Angle	Slope	Post-Treatment Coordinates X	Y	Z
3	25.7	16.5	3.3	-10°	0°	0°	25.9	15.7	4.6
4	23.9	24.5	2.6	-7°	0°	0°	23.9	25.1	3.7
5	22.1	30.9	2.7	-7°	0°	0°	21.6	31.8	4.0
6	19.4	38.9	5.4	0°	0°	0°	17.0	39.1	3.9
7	11.4	42.9	3.2	0°	8°	0°	11.5	44.2	3.4
8	5.2	45.8	3.9	14°	5°	0°	4.5	48.8	4.1
9	3.7	45.3	4.0	14°	5°	0°	4.2	48.9	4.5
10	10.0	43.1	4.7	0°	8°	0°	12.2	44.9	3.4
11	19.4	38.6	6.0	0°	0°	0°	17.7	38.2	4.7
12	21.8	31.1	4.4	-7°	0°	0°	22.4	30.8	3.8
13	24.0	24.7	3.8	-7°	0°	0°	24.7	23.1	3.0
14	25.2	16.2	4.1	-10°	0°	0°	26.9	14.6	4.7

TABLE 2

Tooth	X-axis	Directional Forces (Newtons) Y-axis	Z-axis	Angle Moment (N-m)	Rotational Moment
3	0.398 to 1.432	0.066 to 0.217	0.286 to 0.426	-0.016 to -0.006	-0.246 to -0.075
4	-2.557 to -0.864	-0.500 to -0.151	-1.528 to -1.093	-0.043 to -0.025	-0.089 to 0.001
5	1.928 to 4.796	0..439 to 1.275	1.341 to 1.960	0.017 to 0.059	-0.361 to -0.089
6	-5.235 to -1.970	-3.928 to -1.261	-0.004 to 0.288	0.159 to 0.399	-0.024 to 0.054
7	-0.072 to 0.063	-1.889 to -0.581	-1.613 to -1.209	0.027 to 0.204	0.436 to 1.412
8	0.445 to 1.636	1.924 to 5.522	2.819 to 3.875	0.338 to 0.621	0.122 to 0.525
9	-0.912 to -0.279	0.133 to 1.206	-3.214 to -2.029	0.432 to 0.691	-0.428 to -0.170
10	0.378 to 0.698	-0.613 to -0.159	0.994 to 1.363	0.160 to 0.245	-0.431 to -0.128
11	1.700 to 3.415	-1.679 to -0.583	-3.609 to -2.393	-0.034 to 0.029	0.224 to 0.259
12	-2.162 to -1.522	0.076 to 0.096	2.320 to 3.517	-0.256 to -0.139	0.334 to 0.669
13	-1.325 to -0.405	0.106 to 0.355	-1.751 to -0.286	0.024 to 0.0439	0.013 to 0.095
14	0.128 to 0.285	-0.040 to -0.012	0.186 to 0.276	-0.027 to -0.013	-0.046 to -0.014

Table 1
Spatial (global) pre- and post-movement abutment locations with appliance bracket presettings.

Table 2
Force ranges globally produced using nickel-titanium alloy archwire of varying dimensions (0.016 x 0.025 to 0.024 x 0.025 in).

TABLE 3

Wire Dimension (sq. in.)	TOOTH 3	4	5	6	7	8	9	10	11	12	13	14
0.016 x 0.025	0.21	1.28	2.77	2.98	1.07	2.05	3.43	1.33	2.27	2.37	1.54	0.29
0.017 x 0.025	0.23	1.37	2.94	3.22	1.12	2.20	3.75	1.42	2.72	2.66	1.60	0.57
0.018 x 0.025	0.26	1.47	3.12	3.48	1.19	2.37	4.12	1.55	3.15	2.97	1.69	0.67
0.019 x 0.025	0.27	1.57	3.29	3.74	1.24	2.53	4.48	1.68	3.61	3.30	1.86	0.78
0.020 x 0.025	0.30	1.68	3.46	4.02	1.31	2.71	4.90	1.80	4.12	3.66	2.05	0.89
0.022 x 0.025	0.34	1.93	3.80	4.60	1.46	3.09	5.82	2.12	5.26	4.44	2.46	1.10
0.024 x 0.025	0.39	2.58	4.11	5.22	1.64	3.53	6.92	2.47	6.53	5.32	3.00	1.50
Median Forces	0.30	1.93	3.44	4.10	1.35	2.79	5.17	1.90	4.40	3.84	2.27	0.89

Table 3
Net forces produced on each abutment x changes in archwire dimensions in Newtons (1 Newton = 102 grams/force).

eral abutments were tipped, net forces on such teeth might still be expected to reflect active abutment stresses in this direction. The wire dimensions used in this study were varied to demonstrate this effect on forces applied to the abutment teeth (Table 3). Maintaining preset bracket angles and inclinations would significantly alter the magnitude of net forces while the direction of those forces is largely unaffected. Conversely, varying the inclination or angulation of the brackets while maintaining wire size would be expected to alter both the direction and magnitude of net forces. Where wire size changes are undesirable, the use of a wire with a different elastic modulus could be employed. For example, when this same study was repeated with stainless steel (Permachrome) archwire, all net forces were notably increased (data not shown).

In this study, only initial forces were measured. One can assume that immediately after appliance placement, these forces would be somewhat dissipated by interligamentary movement allowing for limited wire relaxation. However, if one examines the forces generated on tooth #9 (Table 3), it is evident that extreme forces would be generated at wire dimensions of 0.018 x 0.025 in or more even with some anticipated relaxation of initial forces. This data can also be calculated for any point on the appliance wire given the initial and proposed (final) treatment coordinates.

The effect of archwire curvature to the straight beam element equation was not studied in this work. This aspect can be addressed with the proposed technique by introducing more than one beam element between adjacent abutments with coordinates of the common joint measured from the pretorqued-preangulated orthodontic appliance. The significance of such an effect together with more evolved modeling of the abutments (more than two wire contact points) should be further investigated in conjunction with a large deformation linear theory and FEM analysis. While the model utilized in this pilot study presents some limitations, it is capable of providing a feel for the range of force magnitudes anticipated to develop on the appliance and abutments thus setting the basis for more elaborate FEM procedures. At the joints of each beam element, the local force magnitude and direction can readily be evaluated from the components of the net forces through FEM. Additional dynamic force interactions might also be important in some cases when considering occlusal relationships with the mandibular arch. Such additional considerations could modify the actual forces exerted on the abutments. The influence of intra-arch interproximal contact points and the effects of buttressing might also be a point for concern when considering longitudinal abutment stresses and directions of movement which deviate from FEM predictions.

Future studies are planned to delineate the effects of these interactions utilizing the FEM procedure. An examination of initial force relaxation allowed by the periodontal tissues is another area requiring further study. Modeling of the periodontal membrane and supporting alveolar bone has been studied with FEM.[14] Since the resistance to tooth movement exerted by periodontal supporting tissues is known to vary considerably due to endogenous factors such as age, sex, race, etc., this continues to be a major consideration in appliance selection, design, and treatment programming. Future development of this research will require that the influence of these variables be approximated and provided as input for FEM evaluations of specific clinical case studies. Adaptation of such a model coupled with an appliance analysis similar to the simulated case described in this study could produce a powerful utility for predicting general responses to orthodontic therapeutic modes.

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References

1. Lavernia, C.J., Cook, S.D., Weinstein, A.M., Klawitter, J.J.: The influence of the bone-implant stiffness on stress profiles surrounding Al 203 and carbon dental implants. Ann. Biomed. Eng., 10:129, 1982.
2. Anusavice, K.J., Dehoff, P.H., Fairhurst, C.W.: Comparative evaluation of ceramic-metal bond tests using finite element stress analysis. J. Dent. Res., 59:608, 1980.
3. Takahashi, N., Kitagami, T., Komori, T.: Effects of pin-hole position on stress distribution and interpulpal temperatures in horizontal non-parallel pin restorations. J. Dent. Res., 58:2085, 1979.
4. Wright, K., Yettram, A.: Finite element stress analysis of a Class I amalgam subjected to setting and thermal expansion. J. Dent. Res., 57:715, 1978.
5. Farah, J.W., Craig, R.G.: Finite element stress analysis of a restored axiommetric first molar. J. Dent. Res., 53:859, 1974.
6. Takahashi, N., Kitagami, T., Komori, T.: Behavior of teeth under loading conditions with finite element method. J. Oral Rehab., 7:453, 1980.
7. Tanne, K., Sakuda, M.: Initial stress induced in the periodontal tissue at the time of the application of various types of orthodontic force — 3D analysis of means of finite element method. J. Osaka Univ. Bull., 23:143, 1983.
8. Hocevar, R.A., Understanding, planning, and managing tooth movement orthodontic force system theory. Am. J. Orthod., 80:457, 1981.
9. Williams, K.R., Edmundson, J.T.: Orthodontic movement analysed by the finite element method. Biomat., 5:347, 1984.
10. Tanne K., Sakuda, M.: A dynamic analysis of stress in the tooth and its supporting structures: the use of the finite element method as numerical analysis. Nippon Kyosei Shiga Gakkai Zasshi, 38:372, 1979.
11. Wright K.W., Yettram, A.L.: An analytical investigation into possible mechanical causes of bone remodeling. J. Biomed. Eng., 1:41, 1979.
12. Zienkiewicz, O.C.: The finite element method in engineering science. London, McGraw-Hill Book Co., 1974.
13. Smith, R.J., Burstone, C.J.: Mechanics of tooth movement. Am. J. Orthod., 85:294, 1984.
14. Widera, G.E.O.: Interaction effects among cortical bone, cancellous bone, and periodontal membrane of natural teeth and implants. J. Biomed. Mater. Res. Symp., 7:613, 1976.