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Initial stress produced in the periodontal membrane by orthodontic loads in the presence of varying loss of alveolar bone: a three-dimensional finite element analysis

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SUMMARY The aim of this study was to investigate the stress components (S_1 and S_3) that appear in the periodontal membrane (PDM), when subjected to transverse and vertical loads equal to 1 N. A further aim was to quantify the alteration in stress that occurs as alveolar bone is reduced in height by 1, 2.5, 5, 6.5, and 8 mm, respectively.

Six three-dimensional (3D) finite element models (FEM) of a human maxillary central incisor were designed. The models were of the same configuration except for the alveolar bone height. Special attention was paid to changes of the stress components produced at the cervical, apical, and sub-apical levels.

In the absence of alveolar bone loss, a tipping force of 1 N produced stresses, which reached 0.072 N/mm² at the cervical margin, up to 0.0395 N/mm² at the apex and up to 0.026 N/mm² sub-apically. In the presence of 8 mm of alveolar bone loss, the findings were –0.288, 0.472, and 0.722 N/mm², respectively. Without bone loss, an intruding force of the same magnitude produced stresses of –0.0043, –0.0263, and 0.115 N/mm², respectively, for the same areas and sampling points. In the presence of 8 mm of alveolar bone loss the findings were –0.019, –0.043, and 0.185 N/mm² for intrusive movement.

The results showed that alveolar bone loss caused increased stress production under the same load compared with healthy bone support (without alveolar bone resorption). Tipping movements resulted in an increased level of stress at the cervical margin of the PDM in all sampling points and at all stages of alveolar bone loss. These increased stress components were found to be at the sub-apical and apical levels for intrusive movement.

Introduction

Several studies have tried to model the reactions of a tooth and its supporting tissues when loaded with an orthodontic force. However, each method of study has shortcomings:

1. *Photo-elastic stress analysis*: this method needs a well-equipped laboratory and models that should be prepared and used in a specific instrument (Caputo *et al.*, 1974; Aird *et al.*, 1988; Standlee and Caputo, 1988).
2. *Animal studies*: these deal with living tissue and its responses, but it is not possible to correlate all the results with human tissue. In other words, true morphological reflection of the human supporting tissues is not practical (Oppenheim, 1911; Reitan, 1960, 1964; Fortin, 1971).
3. *Mathematical models representing the in vivo situation*: in these models, the shape, form, and function of the concerned process are represented in a form that is compatible with mathematical equations (Hay, 1939; Steyn *et al.*, 1978).
4. *Laser holography*: this procedure needs expensive instruments, although it shows the stress at the surface of the body under study (Burstone and Pryputniewicz, 1980).
5. *Finite element analysis*: this is a numerical form of analysis that allows stresses and displacements to be identified. It involves

discretization of the continuum (dividing the structure of interest) into a number of elements (Desai and Abel, 1972; Reddy, 1993). Discretization is a theoretical subdivision of the structure while it remains its continuity. The problem is solved for each element and then harmonized to reach an answer representative of the entire system. This method has proved effective in many dental fields, such as simulation of tooth movement and optimization of orthodontic mechanics (Farah *et al.*, 1973; Farah and Craig, 1974; Yettram *et al.*, 1977; Tanne *et al.*, 1988, 1991; Wilson *et al.*, 1990; Andersen *et al.*, 1991; McGuinness *et al.*, 1991, 1992; Hart *et al.*, 1992; Koolstra and Van Eijden, 1992; Cobo *et al.*, 1993; Tanne and Matsubara, 1996; Puente *et al.*, 1996; Geramy, 2000a,b,c).

The implications of a reduction of alveolar bone have been assessed from the point of view of change in the centre of resistance (Geramy, 2000a). Although 0.017 mm/year of bone resorption can be considered quite normal (Corn and Mark, 1989), increased resorption can be detected in the patients referred for orthodontic treatment.

Melsen (1991) suggested applying a mild intrusive force in the treatment of adult patients with reduced bone height. On the other hand, there are authors who believe there is an increased risk of root resorption in adult patients when large orthodontic forces are applied to produce continuous bodily (Thilander, 1985) and intrusive movement (Thilander, 1985; Lu *et al.*, 1999).

Maxillary (Ketcham, 1929) and mandibular incisors (Goldson and Henrikson, 1975) have been shown to be more frequently involved than other teeth in apical root resorption. These are two areas of interest in incisor retraction and intrusion in orthodontic treatment.

Apical root resorption is a phenomenon that occurs consequent to some orthodontic treatment. No difference has been found to exist between fixed orthodontic techniques (Beck and Harris, 1994) or between sexes (Spurrier *et al.*, 1990; Beck and Harris, 1994). Without considering the manner of force application, Lu *et al.* (1999) demonstrated the higher activity of resorption in the apical root region than in the inter-radicular area. The occurrence of apical root resorption cannot be predicted on the basis of the morphology of the facial and dento-alveolar structures (Taithongchai *et al.*, 1996). Furthermore, although Costopoulos and Nanda (1996) have shown that it is impossible to correlate the amount of root resorption with the degree of intrusion, the significant differences between a control and intrusion group can be determined statistically.

Resorption has been reported (Spurrier *et al.*, 1990) to be less frequent and less severe in endodontically-treated incisors than in vital (control) teeth. There is evidence that orthodontic root resorption is frequently preceded by hyalinization of the periodontal membrane (PDM) (Rygh, 1977). Kurol and Owman-Moll (1998) have shown hyalinized areas opposite an intact root surface (54 per cent) or close to an area of root resorption (45 per cent). Lee (1965), in a study to find the so-called optimum stress for tooth movement (force on the root or the surrounding bone-PDM complex), reported an optimum in the range from 0.0165 to 0.0185 N/mm^2 (1.65–1.85 gf/mm^2) and stated that the actual stress exerted was likely to be slightly higher because of consequent alveolar bone loss. In a recent study, Lee (1996) reported an increased stress value of 0.0197 N/mm^2 (1.97 gf/mm^2) to be optimal for tooth movement.

Stress distribution, hyalinization and root resorption are interrelated events that result from force application. The aims of this study were to quantify stress components produced consequent to force application during orthodontic treatment, in the presence of a healthy tooth support and in subjects with a gradual loss of alveolar bone.

Materials and methods

Six three-dimensional (3D) finite element models (FEM) of a maxillary central incisor were designed. Each model consisted of 556–726 nodes and 310–475 elements (Figure 1) based on Ash's dental anatomy (Ash, 1984) with minor

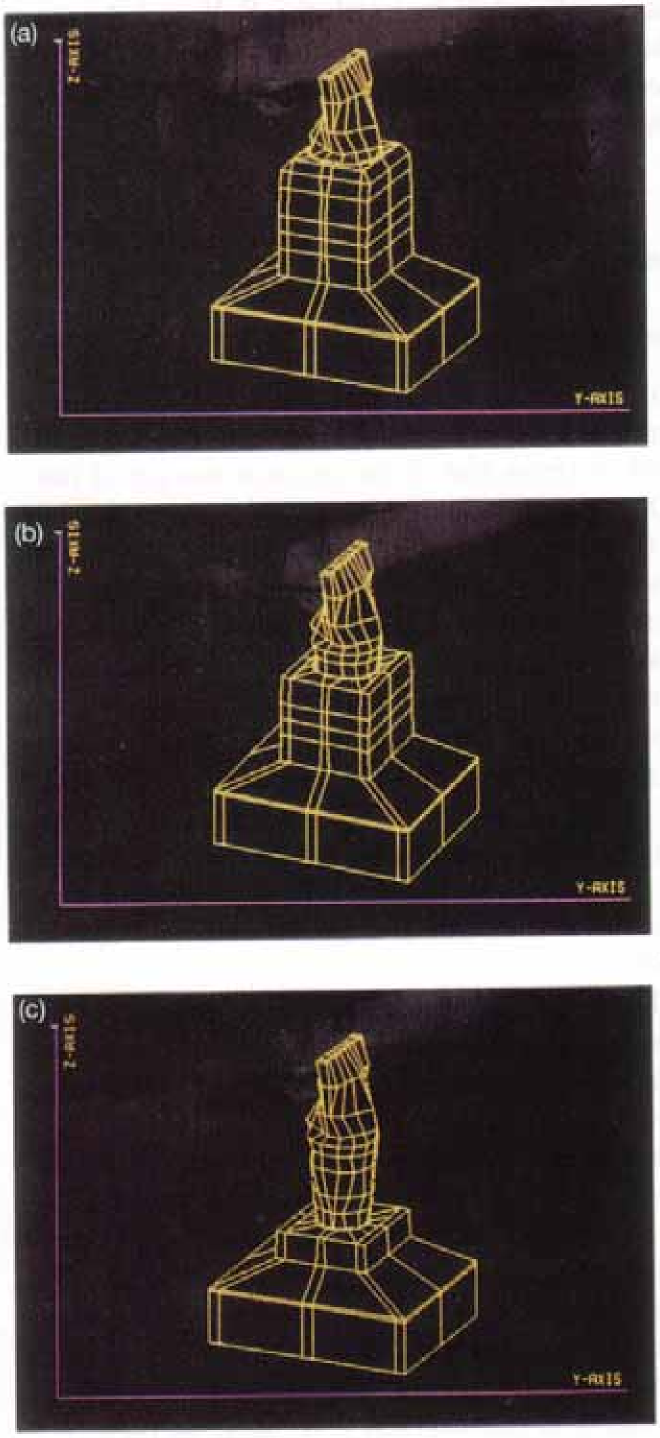

Figure 1 3D models used in the study. (a) Normal alveolar bone height. (b) 2.5 mm of alveolar bone loss. (c) 8 mm of alveolar bone loss.

modifications to obtain the best shape. A 3D brick isoparametric octahedral element was chosen to construct the models. Each model contained a tooth, its PDM, spongy, and cortical bone. Each section was given 14 external nodes to enable appropriate modelling (Figure 2). The

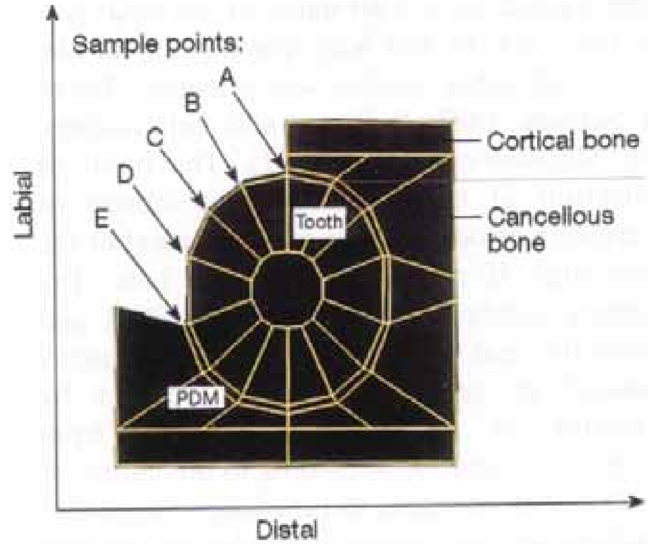

Figure 2 Cross-section of the model at 10.5 mm. Incisal to the apex, showing the relationship of the involved tissues and the sampling points.

root was split into seven levels of varying vertical heights, 1 mm at the cervical, 1.5 mm at the mid-root area, and 2.5 mm for other levels (Figure 1). The use of a mesh with different element sizes provides results that are more accurate where needed and minimizes node numbers in the regions of less importance. According to Coolidge (1937) the PDM is of varying thickness *in vivo* (Table 1). The use of these varying thicknesses makes the models more precise and realistic. The only difference between the models used in this investigation was the height of the alveolar bone, which was assumed to be equal to 13 (normal alveolar bone height), 12, 10.5, 8, 6.5, and 5 mm, respectively.

Table 1 Geometry of the PDM widths in the 3D model according to Coolidge (1937), with different PDM thickness at various locations.

Distance from the alveolar crest (mm)	Distal (mm)	Lingual (mm)	Mesial (mm)	Labial (mm)
13.0	0.25	0.25	0.22	0.25
10.5	0.18	0.22	0.20	0.22
8.0	0.15	0.20	0.17	0.20
6.5	0.14	0.18	0.16	0.18
5.0	0.15	0.20	0.17	0.20
2.5	0.18	0.22	0.20	0.22
0	0.19	0.24	0.21	0.24

The applied force at all stages of this study was 1 N (98.1 gf). In this way, comparison of the results with other studies was possible (Tanne and Sakuda, 1983; Williams and Edmundson, 1984; McGuinness *et al.*, 1991). The point of application of the force was the centre of the crown mesiodistally, i.e. 5.5 mm apical to the incisal edge, 16 mm from the model base. The boundary condition is important in FEM and reflects the real situation of the displacements produced at each node. It is defined by restraining or releasing some nodes from movement or rotation according to the nature of the 3D model, its shape, its function, or any other consideration that makes it similar to the real condition. Brick elements may experience displacements along the X, Y, and Z planes, which can be limited by defining restraints and constraints. The boundary condition of these models is defined so that all the movements at the base of the model are restrained. This manner of restraining prevents the model from any rigid body motion, while the load is acting. The analyses were performed on a Pentium II personal computer by Super sap Version 7.1 (Algor Interactive Systems, Inc., Pittsburgh, USA). The material properties were derived from other investigations (Table 2). In this study, Poisson's ratio of the PDM was assumed to be 0.49, which gives an incompressible nature to it and explains its biological characteristics rather accurately (Table 2).

The direction of applied force was as follows: linguo-labially to tip the tooth labially (F = 1 N) and inciso-gingivally for intrusive movement (F = 1 N) and a couple to neutralize the tipping tendency of the applied force.

The lack of any distance from the longitudinal tooth axis in the case of linguo-labial force application avoided any tendency for rotation.

Five different points positioned on the periphery of the root (A, B, C, D, and E) were selected (Figure 2).

Among the criteria used to evaluate the structure under loading to judge the stress at certain points, the so-called maximum (S_1) and minimum (S_3) principal stresses were used. These stresses occur in the plane of zero shear stress, perpendicular to each other but not necessarily coincident with the X, Y, and Z-axes. The results were evaluated for PDM, because tooth movement is essentially a PDM-based phenomenon.

Results

Tipping movement

Normal alveolar bone height. In the case of linguo-labially directed forces, the increased level of stress was at the cervical margin of the PDM at all sampling points, –0.072 N/mm² for A; –0.062 N/mm² for B; 0.0016 N/mm² for C; 0.052 N/mm² for D; and 0.055 N/mm² for E (Table 3). Moving from the cervical or the apical region towards the centre of rotation decreased the stress level (Figure 3). There was a change of the sign in stress (compressive versus tensile), and an increase of stress in sampling points A, B, D, and E. However, point C did not follow the same pattern. This can be explained by the direction of tooth movement and its location. Stress signs were the same at the apical and sub-apical level.

The maximum stress at the sub-apical level was approximately 37.5 per cent of the cervical stress. In the presence of tipping loads, point C showed the lowest amount of stress of all the sampling points at the cervical margin (Table 3). *Alveolar bone loss.* The highest stress values (S_1 and S_3) were found at the cervical level (Table 3). Maximum principal stress (S_1) increased gradually with an increase of alveolar bone resorption (Figure 3). Assessment of the stress values revealed that the least values of S_1 and S_3 at the cervical level were at point C, the absolute values increasing towards points A and E (Figure 4). The same pattern held for S_3.

Careful assessment of Table 3 revealed two main results:

1. There was a change of the stress sign along the root surface in all sampling points except for 'C'.
2. Stresses produced at the apical and the sub-apical level were of the same sign.

Table 2 A review of the physical properties used by researchers and those in the present study. All the values are in N/mm².

	Tooth		PDM		Cancellous bone		Cortical bone	
	Young's modulus	Poisson's ratio	Young's modulus	Poisson's ratio	Young's modulus	Poisson's ratio	Young's modulus	Poisson's ratio
Takahashi *et al.* (1978)			9.8	0.45				
Craig and Farah (1978)			3.45	0.45				
Wright and Yettram (1979)			0.08 axial, 0.12 oblique 0.16 lat. loading	0.3	13,800	0.3	137,900 stiffer dir. 69,000 lat. dir.	0.3 & 0.15
Peters *et al.* (1983)			3.45	0.45				
Farah *et al.* (1988)			6.9	0.45	25,000	0.3	100,000	0.3
Maeda and Wood (1989)					15,000	0.3	100,000	0.3
Richter *et al.* (1990)					20,000	0.2	200,000	0.3
Williams and Edmundson (1984)			0.5–99.9	0–0.45	13,700	0.38	340,000	0.26
Andersen *et al.* (1991)			0.07	0.49	Mandibular segments 7930 Intact mandible 11,000	0.3	Mandible 137,000	0.3
Tanne *et al.* (1989)	20,700	0.3			8000	0.3	137,000	0.3
Tanne *et al.* (1991) Cobo *et al.* (1993)	20,000	0.15	0.68	0.49	Alv bone 14,000	0.15		
Tanne *et al.* (1998)	19,600	0.3	0.666	0.49	Alv bone 13,700	0.3		
Present study	20,300	0.3	0.667	0.49	13,700	0.38	340,000	0.26

Table 3 Stress components of the PDM at different root levels and sampling points with diverse degrees of alveolar bone loss when subjected to a tipping force (F = 1 N).

Root level	Sampling point	Bone loss = 0 mm		Bone loss = 1 mm		Bone loss = 2.5 mm		Bone loss = 5 mm		Bone loss = 6.5 mm		Bone loss = 8 mm	
		S_1	S_3	S_1	S_3	S_1	S_3	S_1	S_3	S_1	S_3	S_1	S_3
12–13	A	–0.072	–0.075										
	B	–0.062	–0.065										
	C	0.0016	–0.0015										
	D	0.052	0.050										
	E	0.055	0.053										
10.5–12	A	–0.055	–0.058	–0.084	–0.088								
	B	–0.039	–0.042	–0.061	–0.064								
	C	0.0013	–0.0011	0.0021	–0.0015								
	D	0.032	0.030	0.049	0.046								
	E	0.039	0.037	0.060	0.057								
8–10.5	A	–0.034	–0.037	–0.055	–0.058	–0.101	–0.107						
	B	–0.025	–0.026	–0.039	–0.042	–0.072	–0.076						
	C	8.4e–4	–7.8e–4	0.0014	–0.0012	0.0023	–0.0025						
	D	0.027	0.025	0.043	0.040	0.077	0.073						
	E	0.037	0.035	0.059	0.056	0.109	0.103						
6.5–8	A	–0.0015	–0.0033	–0.0035	–0.006	–0.0089	–0.013	–0.048	–0.062				
	B	–0.0025	–0.003	–0.0056	–0.0062	–0.014	–0.016	–0.077	–0.081				
	C	4.8e–4	1.4e–4	0.0011	2.8e–4	0.0012	–8e–4	0.005	–0.006				
	D	0.027	0.0039	0.0099	0.0084	0.023	0.020	0.108	0.112				
	E	0.037	0.0097	0.021	0.017	0.046	0.042	0.217	0.204				
5–6.5	A	0.0044	0.0027	0.0038	0.0015	–0.0022	–0.006	–0.062	–0.072	–0.222	–0.246		
	B	0.0049	0.0041	0.0051	0.004	0.0019	9.6e–5	–0.037	–0.042	–0.148	–0.161		
	C	–5.7e–4	–9.7e–4	–7.4e–4	–1.2e–3	–5.7e–4	–0.001	0.0028	0.0013	0.015	0.0082		
	D	–0.0036	–0.0046	–0.0027	–0.0042	0.0013	–5.9e–4	0.051	0.046	0.186	0.174		
	E	–0.0022	–0.0040	–2.7e–4	–2.9e–3	0.0063	0.0024	0.073	0.062	0.248	0.229		
2.5–5	A	0.0205	0.0193	0.028	0.025	0.036	0.034	0.046	0.041	0.0103	0.0101	–0.288	–0.316
	B	0.014	0.013	0.018	0.017	0.024	0.022	0.031	0.027	0.0072	4.4e–4	–0.191	–0.211
	C	0.001	–3.6e–4	0.0016	–2.3e–4	0.0016	–9.7e–4	0.0025	–0.0014	0.002	–0.0014	3.9e–4	–8.5e–3
	D	–0.012	–0.013	–0.016	–0.017	–0.021	–0.023	–0.026	–0.029	0.001	–0.0058	0.213	0.193
	E	–0.019	–0.020	–0.026	–0.027	–0.035	–0.037	–0.042	–0.048	–0.0015	–0.0114	0.330	0.30
0–2.5*	A	0.039	0.037	0.055	0.052	0.080	0.076	0.167	0.158	0.265	0.250	0.472	0.433
	B	0.028	0.026	0.040	0.037	0.057	0.053	0.119	0.111	0.190	0.175	0.338	0.307
	C	0.0015	–6.6e–4	0.0026	–4.7e–4	0.0025	–0.0021	0.0053	–0.0045	0.0086	–0.0078	0.015	–0.018
	D	–0.025	–0.027	–0.034	–0.037	–0.051	–0.055	–0.107	–0.116	–0.168	–0.183	–0.294	–0.325
	E	–0.037	–0.039	–0.051	–0.054	–0.077	–0.018	–0.160	–0.170	–0.251	–0.0268	–0.44	–0.475
–0.25–0**	A	0.027	0.022	0.036	0.030	0.062	0.052	0.149	0.127	0.28	0.24	0.722	0.633
	B	0.024	0.020	0.033	0.028	0.055	0.026	0.130	0.111	0.24	0.207	0.600	0.522
	C	3.8e–4	–9.2e–4	2e–4	–0.0016	0.0015	–0.0013	0.0035	–0.0029	0.0065	–0.0047	0.017	–0.0088
	D	–0.022	–0.026	–0.033	–0.038	–0.047	–0.055	–0.111	–0.130	–0.206	–0.240	–0.511	–0.589
	E	–0.025	–0.029	–0.038	–0.044	–0.053	–0.062	–0.129	–0.151	–0.245	–0.284	–0.632	–0.722

*Apical area; **sub-apical area.

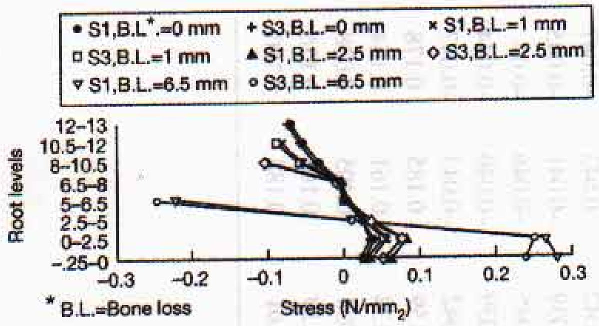

Figure 3 Alteration of S_1 and S_3 in tipping movement due to alveolar bone loss. The S_1 and S_3 patterns for each alveolar bone loss is almost the same. Increase of the cervical stress at each phase of the study is obvious.

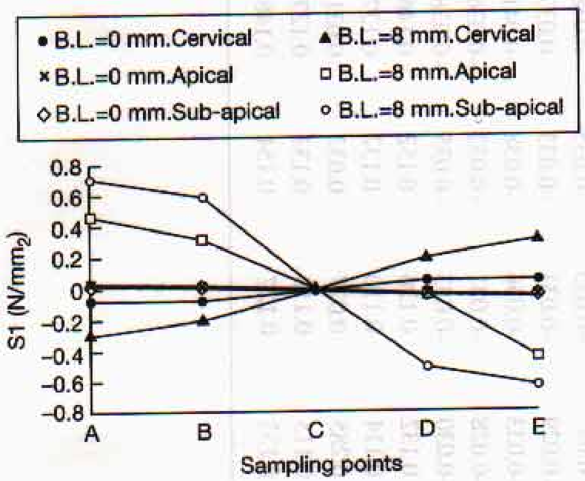

Figure 4 Variation of S_1 in tipping movement at different sampling points with diverse degrees of alveolar bone loss. The difference between the stresses produced at the sampling points is minimal at the normal alveolar bone height. The more alveolar bone loss occurred, the greater the difference between sampling point stresses. There is a change of the stress sign along the root surface.

Pure intrusion

Normal alveolar bone height. The highest level of stress was found at the sub-apical level, with the next highest at the apical level (Table 4). In the case of S_1 the values were 0.115 N/mm² at point A of the sub-apical level, –0.026 N/mm² at the same sampling point of the apical level, and –0.0043 N/mm² at the cervical level. There was no change in stress sign (tensile versus compressive) along the root surface (Figure 5). The PDM underwent tensile stress along the root surface and the sub-apical level experienced compression.

Alveolar bone loss. Assessment of the whole findings (Table 4) revealed the highest stresses at the sub-apical level (Figure 5). The principal stress components (S_1 and S_3) at the apical and sub-apical levels show that stresses at these two levels were of a differing nature (tensile versus compressive). Comparison of principal stress components (S_1 and S_3), at sampling points A to E, revealed that point C had the lowest stress level (Figure 6). The sub-apical level exhibited the highest values of S_1 and S_3 relative to other areas. There was an increase of S_1 with alveolar bone loss (Figure 6). The same pattern holds for S_3.

Discussion

It should be noted that in a numerical method such as FEM, the results relate directly to the input data. Consequently, discrepancies in the accepted values for the physical properties of the supporting tissues can alter the results. Variation in the values reported for the physical properties of the tissues in different studies has been carefully examined by McGuinness *et al.* (1992) and Tanne *et al.* (1998).

Tanne *et al.* (1987), in a 3D FEM study, reported a cervical margin stress of 0.012 N/mm² (1.2 gf/mm²) when a lingually directed tipping force of 1 N was applied to the centre of a mandibular premolar model. This value is almost the same as that reported by Tanne and Sakuda (1983), i.e. that a stress of 0.01 N/mm² occurs at the cervical level in response to a 1 N force, applied mesiodistally on the labial surface of a tooth. McGuinness *et al.* (1991) reported a stress value of 0.132 N/mm² at the cervical margin for a mesiodistally-directed tipping force of 1 N applied to the centre of the tooth, while the stress value at the apex was 0.002 N/mm². McGuinness *et al.* (1992) studied the maximum principal cervical stress (S_1) and reported a value of 0.072 N/mm², while that at the apex was 0.0038 N/mm².

The rotational tendency accompanying a tipping movement and the manner of its elimination has not been discussed. The result of this study for S_1 at sampling point A was 0.072 N/mm² at the cervical, 0.0395 N/mm² at the apical, and 0.0268 N/mm² at the sub-apical levels.

Table 4 Stress components (S_1 and S_3) of the PDM in sampling points at different levels with diverse degrees of alveolar bone loss when subjected to an intrusive loading (F = 1 N).

Root level	Sampling point	Bone loss = 0 mm		Bone loss = 1 mm		Bone loss = 2.5 mm		Bone loss = 5 mm		Bone loss = 6.5 mm		Bone loss = 8 mm	
		S_1*	S_3	S_1	S_3	S_1	S_3	S_1	S_3	S_1	S_3	S_1	S_3
Cervical	A	−0.0043	−0.0077	−0.0089	−0.0124	−0.0054	−0.0098	0.0012	−0.0042	−0.018	−0.024	−0.019	−0.023
	B	−0.012	−0.156	−0.0036	−0.007	−0.011	−0.015	−0.0011	−0.004	−0.008	−0.014	−0.016	−0.02
	C	−0.0065	−0.0101	−0.0109	−0.0147	−0.002	−0.007	−0.018	−0.025	0.0087	0.0045	−0.024	−0.029
	D	−0.016	−0.019	−6.8e−4	−0.0032	−0.011	−0.015	−0.0024	−0.0066	−0.0131	−0.0166	−0.016	−0.0198
	E	−0.009	−0.012	−0.0046	−0.0017	−0.0055	−0.010	−0.012	−0.017	−0.0181	−0.0242	0.019	−0.023
Apical	A	−0.026	−0.029	−0.028	−0.031	−0.03	−0.033	−0.035	−0.038	−0.038	−0.042	−0.043	−0.047
	B	−0.025	−0.028	−0.027	−0.03	−0.029	−0.031	−0.033	−0.037	−0.036	−0.039	−0.041	−0.045
	C	−0.028	−0.031	−0.03	−0.033	−0.033	−0.036	−0.038	−0.041	−0.041	−0.045	−0.046	−0.05
	D	−0.025	−0.027	−0.0259	−0.0286	−0.028	−0.031	−0.032	−0.036	−0.036	−0.039	−0.040	−0.044
	E	−0.026	−0.029	−0.0276	−0.0304	−0.030	−0.033	−0.035	−0.038	−0.038	−0.042	−0.043	−0.047
Sub-apical	A	0.115	0.110	0.122	0.117	0.132	0.126	0.152	0.146	0.166	0.159	0.185	0.178
	B	0.099	0.096	0.106	0.101	0.114	0.110	0.132	0.127	0.144	0.138	0.161	0.154
	C	0.025	0.024	0.026	0.025	0.285	0.027	0.033	0.031	0.0361	0.034	0.405	0.038
	D	0.1003	0.096	0.107	0.103	0.115	0.110	0.133	0.127	0.145	0.139	0.162	0.156
	E	0.116	0.112	0.124	0.119	0.133	0.127	0.154	0.148	0.168	0.161	0.188	0.180

*N/mm².

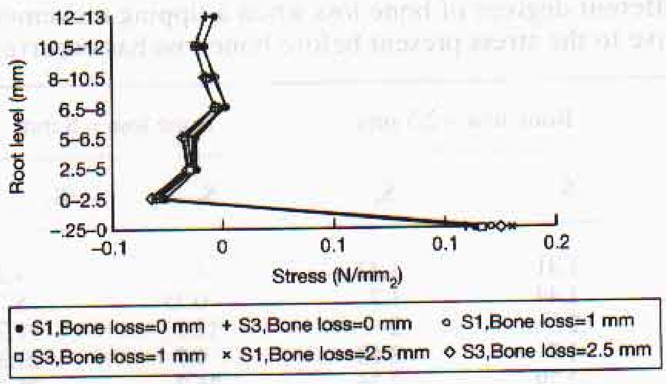

Figure 5 Alteration of S_1 and S_3 with diverse degrees of alveolar bone loss when loaded by an intrusive force. There was no significant change in the stress state due to alveolar bone loss. (On the contrary, a tipping movement was influenced significantly by alveolar bone loss.)

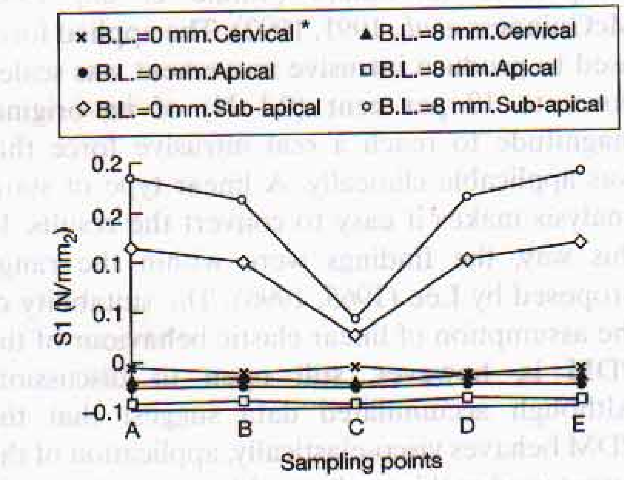

Figure 6 Variation of S_1 produced by an intrusive movement at different sampling points with diverse degrees of alveolar bone loss. The stress produced at the sub-apical level was significantly influenced by alveolar bone loss. There were minimum changes in the stresses produced at the apical and cervical levels.

The values obtained in this study are higher than those of Tanne and Sakuda (1983), Tanne *et al.* (1991), McGuinness *et al.* (1992), and Puente *et al.* (1996). The result for the apical stress at point A in the normal situation (0.039 N/mm²) is well below the value of 0.05 N/mm² (5 gf/mm²) obtained by Cobo *et al.* (1993).

Alveolar bone loss causes an increase in the stress values at all sampling points. The maximum principal stress at the apical level, in the presence of a tipping movement and 8 mm of alveolar bone loss, was –0.288 N/mm², which is almost in agreement with the value obtained by Cobo *et al.* (1993), i.e. 0.364 N/mm² (36.36 gf/mm²). The stress sign differs due to a different force direction. Assuming a linear relationship between the applied force and the resulting stress, and the maximum stress suggested by Lee (1965, 1996), the optimum-tipping force would be approximately 0.36 N (36 gf).

The application of higher levels of force would result in somewhat higher stresses than suggested, which means anticipating a hyalinization area at the cervical level. An important consideration is the stress produced at the sub-apical level, the thin level under the apical part of the apex, which is just as important as other parts of the PDM. The finding of a tipping movement in the normal situation (first column Table 3) indicates that almost 37.5 per cent of the stress (S_1) at the cervical level will be produced on the labial side of the sub-apical level. Sub-apical stress (S_1) increases up to 2.5× at point A, 3.1× in the mesiolabial direction (point B), and up to 43× at point C relative to cervical stress in the presence of 8 mm alveolar bone loss. The stress increment ratios for S_1 in subsequent steps of alveolar bone loss are shown in Table 5.

The highest ratio of stress increase, due to reduced alveolar support, occurs in the sub-apical level. The comparison of the ratio of sub-apical to cervical area stress (first and last columns Table 3) shows that the ratio varies from 1:0.375 for normal alveolar bone height to approximately 2:1 for 8 mm of alveolar bone loss. The stress occurring in the cervical level at point A during intrusive tooth movement in the model without alveolar bone loss was –0.0043 N/mm²; which is in agreement with the value of 0.0046 N/mm² reported by Wilson *et al.* (1994) for the same conditions. However, the values for apical stress reported by Wilson *et al.* (1994) differ from the findings of this study. Their model showed the highest stresses at the cervical level, but in the present study the highest stress level (S_1) was exhibited at the apical and the sub-apical levels. The current findings (Table 4)

Table 5 Increase in stress for points A and C with different degrees of bone loss when a tipping movement is induced. The increases are expressed as a ratio relative to the stress present before bone loss has occurred.

Sampling point		Bone loss = 0 mm		Bone loss = 2.5 mm		Bone loss = 8 mm	
		S_1	S_3	S_1	S_3	S_1	S_3
Cervical	A	1	1	1.41	1.42	4	4.21
	C	1	1	1.44	1.7	0.25	5.7
Apical	A	1	1	2	2	12.1	11.7
	C	1	1	1.7	3.18	9.7	27.3
Sub-apical	A	1	1	2.29	2.36	26.7	28.77
	C	1	1	3.94	1.41	44.12	9.56

indicate that the stress (S_1) produced at the sub-apical level is about 26 times that at the cervical level. The comparable ratio for apical stress (S_1) is about 6. In spite of the production of the highest stresses at the sub-apical level, the highest ratio of the stress increment with decreasing alveolar bone height is at the cervical level, which is up to 4.5× more than normal at sampling point A (Table 6). This finding is approximately double that for the same point at the apical and sub-apical levels. The differences between the present results and those of other authors seem to be due to the following:

1. Different physical properties are used.
2. Different simplification assumptions are adopted in modelling.
3. Original teeth used for modelling have normal variation in dimensions and configurations.

The application of a force of 1 N in all phases of this investigation made the study and results comparable to others (Tanne *et al.*, 1987; McGuinness *et al.*, 1991, 1992). The applied force used to produce intrusive movement was scaled down to 10 per cent (0.1 N) of its original magnitude to reach a real intrusive force that was applicable clinically. A linear type of static analysis makes it easy to convert the results. In this way, the findings were within the range proposed by Lee (1965, 1996). The suitability of the assumption of linear elastic behaviour of the PDM is, however, still open to discussion. Although accumulated data suggest that the PDM behaves visco-elastically, application of the forces used in this study would cause only a small amount of initial tooth movement. It was, therefore, considered reasonable to assume a linear-elastic behaviour for all the materials involved.

Table 6 Increase in stress for points A and C with different degrees of bone loss when an intrusive movement is induced. The increases are expressed as a ratio relative to the stress present before bone loss has occurred.

Sampling points		Bone loss = 0 mm		Bone loss = 2.5 mm		Bone loss = 8 mm	
		S_1	S_3	S_1	S_3	S_1	S_3
Cervical	A	1	1	1.25	1.26	4.42	2.98
	C	1	1	0.31	0.72	3.69	2.86
Apical	A	1	1	Approximately	1.13	Approximately	1.6
	C	1	1				
Sub-apical	A	1	1	Approximately	1.14	Approximately	1.6
	C	1	1				

Conclusions

1. When tipping movements occur in the model, the highest level of stress is found at the cervical level.
2. Alveolar bone resorption caused an increase in the maximum principal stress (S_1) and minimum principal stress (S_3), relative to that found with a normal bone height, of up to 4.19 times at point A of the cervical level, up to 26.6 times at the apical level at point C, and up to 44.1 times at the sub-apical level at point C.
3. Sub-apical stress components (S_1 and S_3) were evaluated as an aspect of stress distribution.
4. The findings in relation to tipping movements indicate that a force equal to 36 gf produces stresses within the limits suggested by Lee (1965). It is obvious that application of higher forces will result in higher stresses, which may cause hyalinization and other side-effects.
5. Application of a light force causes a small amount of initial tooth movement; therefore, it is considered reasonable to assume a linear-elastic behaviour for the materials involved.
6. An intrusive movement causes a stress concentration at the sub-apical and apical levels. (Normal variation of the root configuration may affect this.)
7. Alveolar bone resorption of 8 mm results in an increase of stress (S_1) when the tooth is subjected to a vertical load. The increased ratios are 4.5 times at the cervical level (S_1 at sampling point A) and up to 1.6 times at the apical and sub-apical levels.
8. Scaling the findings for intrusive movement down to 10 per cent (i.e. reaching the stress produced by a force of only 0.1 N), which is an acceptable intrusive force clinically, no stresses (S_1 and S_3) were found higher than the optimum levels proposed by Lee (1965, 1996).

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References

Aird J C, Millett D T, Sharples K 1988 Fracture of polycarbonate brackets—a related photoelastic stress analysis. British Journal of Orthodontics 15: 87–92

Andersen K L, Pedersen K H, Melsen B 1991 Material parameters and stress profile within the periodontal ligament. American Journal of Orthodontics and Dentofacial Orthopedics 99: 427–440

Ash M M 1984 Dental anatomy, physiology and occlusion, 6th edn. W B Saunders, Philadelphia

Beck B W, Harris E F 1994 Apical root resorption in orthodontically treated subjects: analysis of edgewise and light wire mechanics. American Journal of Orthodontics and Dentofacial Orthopedics 105: 350–361

Burstone C J, Pryputniewicz R J 1980 Holographic determination of centers of rotation produced by orthodontic forces. American Journal of Orthodontics 77: 396–409

Caputo A, Chaconas S J, Hayashi R K 1974 Photoelastic visualization of orthodontic forces during canine retraction. American Journal of Orthodontics 65: 250–259

Cobo J, Sicilia A, Argüelles J, Suárez D, Vijande M 1993 Initial stress induced in periodontal tissue with diverse degrees of bone loss by an orthodontic force: tri-dimensional analysis by means of the finite element method. American Journal of Orthodontics and Dentofacial Orthopedics 104: 448–454

Coolidge E D 1937 The thickness of human periodontal membrane. Journal of the American Dental Association and Dental Cosmos 24: 1260–1270

Corn H, Mark M H (eds) 1989 Basic biologic concepts associated with adult orthodontics. In: Atlas of adult orthodontics. Lea & Febiger, Philadelphia

Costopoulos G, Nanda R 1996 An evaluation of root resorption incident to orthodontic intrusion. American Journal of Orthodontics and Dentofacial Orthopedics 109: 543–548

Desai C S, Abel J F 1972 Introduction to the finite element method. Van Nostrand Rheinold Publication Co., New York

Farah J W, Craig R G 1974 Finite element stress analysis of a restored axisymmetric first molar. Journal of Dental Research 53: 859–866

Farah J W, Craig R G, Sikarskie D L 1973 Photoelastic and finite element stress analysis of a restored axisymmetric first molar. Journal of Biomechanics 6: 511–520

Farah J W, Craig R G, Meroueh K A 1988 Finite element analysis of a mandibular model. Journal of Oral Rehabilitation 15: 615–624

Fortin J W 1971 Translation of premolars in the dog by controlling the moment to force ratio on the crown. American Journal of Orthodontics 59: 541–551

Geramy A 2000a Alveolar bone resorption and the center of resistance modification (3-D analysis by means of the finite element method). American Journal of Orthodontics and Dentofacial Orthopedics 117: 399–405

Geramy A 2000b Moment/force ratio and the centre of rotation: 3D analysis by means of the finite element method. Journal of Shiraz University of Medical Sciences Dental School 1: 26–34

Geramy A 2000c The cervical headgear force system: 3D analysis by means of the finite element method. Journal of Shiraz University of Medical Sciences Dental School 2: 21–30

Goldson L, Henrikson C O 1975 Root resorption during Begg treatment: a longitudinal roentgenographic study. American Journal of Orthodontics 68: 55–66

Hart R T, Hennebel V V, Thongpreda N, Van Buskirk W C, Anderson R C 1992 Modelling the biomechanics of the mandible: a three-dimensional finite element study. Journal of Biomechanics 25: 261–286

Hay G E 1939 The equilibrium of a thin compressible membrane with application to the periodontal membrane. Canadian Journal of Research 17: 123–140

Ketcham A H 1929 A progress report of an investigation of apical root resorption of vital permanent teeth. International Journal of Orthodontics 15: 310–328

Koolstra J H, Van Eijden T M G J 1992 Application and validation of a three-dimensional mathematical model of the human masticatory system *in vivo*. Journal of Biomechanics 25: 175–187

Kurol J, Owman-Moll P 1998 Hyalinisation and root resorption during early orthodontic tooth movement in adolescents. Angle Orthodontist 68: 161–165

Lee B W 1965 Relationship between tooth movement rates and estimated pressure applied. Journal of Dental Research 44: 1053

Lee B W 1996 The force requirement for tooth movement. Part III: the pressure hypothesis tested. Australian Orthodontic Journal 14: 93–97

Lu L H, Lee K, Imoto S, Kyomen S, Tanne K 1999 Histologic and histochemical quantification of root resorption incident to the application of intrusive force to rat molars. European Journal of Orthodontics 21: 57–63

Maeda J, Wood W W 1989 Finite element method simulation of bone resorption beneath a complete denture. Journal of Dental Research 68: 1370–1373

McGuinness N J, Wilson A N, Jones M L, Middleton J 1991 A stress analysis of the periodontal ligament under various loadings. European Journal of Orthodontics 13: 231–242

McGuinness N, Wilson A N, Jones M, Middleton J, Robertson N R 1992 Stresses induced by edgewise appliances in the periodontal ligament, a finite element study. Angle Orthodontist 62: 15–21

Melsen B (ed.) 1991 Limitations of adult orthodontics. In: Current controversies in orthodontics. Quintessence Publishing Company Ltd, Chicago, pp. 147–180

Oppenheim A 1911 Tissue changes, particularly of the bone incident to tooth movement. International Journal of Orthodontia, Orthodontics and Dentistry for Children 3: 57–58

Peters M C R B, Poort H W, Farah J W, Craig R G 1983 Stress analysis of a tooth restored with a post and core. Journal of Dental Research 62: 760–763

Puente M I, Galbán L, Cobo J M 1996 Initial stress differences between tipping and torque movements. A three dimensional finite element analysis. European Journal of Orthodontics 18: 329–339

Reddy J N 1993 An introduction to the finite element method. McGraw-Hill, New York

Reitan K 1960 Tissue behavior during orthodontic tooth movement. American Journal of Orthodontics 46: 881–900

Reitan K 1964 Effects of force magnitude and direction of tooth movement on different alveolar bone types. Angle Orthodontist 34: 244–255

Richter E J, Orschall B, Jovanovic S A 1990 Dental implant abutment resembling the two-phase tooth mobility. Journal of Biomechanics 23: 297–306

Rygh P 1977 Orthodontic root resorption studied by electron microscopy. Angle Orthodontist 47: 1–16

Spurrier S W, Hall S H, Joondeph D R, Shapiro P A, Reidel R A 1990 A comparison of apical root resorption during orthodontic treatment in endodontically treated and vital teeth. American Journal of Orthodontics and Dentofacial Orthopedics 97: 130–134

Standlee J P, Caputo A A 1988 Load transfers by fixed partial dentures with three abutments. Quintessence International 19: 403–410

Steyn C L, Verwoerd W S, Van der Merwe E J, Fourie O L 1978 Calculation of the position of the axis of rotation when single-rooted teeth are orthodontically tipped. British Journal of Orthodontics 5: 153–156

Taithongchai R, Sookkorn K, Killiany D M 1996 Facial and dentoalveolar structure and the prediction of apical root shortening. American Journal of Orthodontics and Dentofacial Orthopedics 110: 296–302

Takahashi N, Kitagami T, Komori T 1978 Analysis of stress on a fixed partial denture with a blade-vent implant abutment. Journal of Prosthetic Dentistry 40: 186–191

Tanne K, Sakuda M 1983 Initial stress induced in the periodontal tissue at the time of the application of various types of orthodontic force: three-dimensional analysis by means of the finite element method. Journal of Osaka University Dental School 23: 143–171

Tanne K, Matsubara S 1996 Association between the direction of orthopedic headgear force and sutural response in the nasomaxillary complex. Angle Orthodontist 66: 125–130

Tanne K, Sakuda M, Burstone C J 1987 Three-dimensional finite element analysis for stress in the periodontal tissue by the orthodontic forces. American Journal of Orthodontics and Dentofacial Orthopedics 92: 499–505

Tanne K, Koenig H A, Burstone C J 1988 Moment to force ratios and the center of rotation. American Journal of Orthodontics and Dentofacial Orthopedics 94: 426–431

Tanne K, Hiraga J, Sakuda M 1989 Effects of directions of maxillary protraction forces on biomechanical changes in craniofacial complex. European Journal of Orthodontics 11: 382–391

Tanne K, Nagataki I, Inoue Y, Sakuda M, Burstone C J 1991 Patterns of initial tooth displacements associated with various root lengths and alveolar bone heights. American Journal of Orthodontics and Dentofacial Orthopedics 100: 66–71

Tanne K, Yoshida S, Kawata T, Sasaki A, Knox J, Jones M L 1998 An evaluation of the biomechanical response of the tooth and periodontium to orthodontic forces in adolescent and adult subjects. British Journal of Orthodontics 25: 109–115

Thilander B 1985 Indication for orthodontic treatment in adults. In: Thilander B, Rönning O (eds) Introduction to orthodontics. Tandläkarförlaget, Stockholm, p. 237

Williams K R, Edmundson J T 1984 Orthodontic tooth movement analysed by finite element method. Biomaterials 5: 347–351

Wilson A N, Middleton J, McGuinness N, Jones M L 1990 A finite element study of canine retraction with a palatal spring. British Journal of Orthodontics 18: 211–218

Wilson A N, Middleton J, Jones M L, McGuinness N J 1994 The finite element analysis of stress in the periodontal ligament when subject to vertical orthodontic forces. British Journal of Orthodontics 21: 161–167

Wright K W J, Yettram A L 1979 Reactive force distribution for teeth when loaded singly and when used as fixed partial denture abutments. Journal of Prosthetic Dentistry 42: 411–416

Yettram A L, Wright K W J, Houston W J B 1977 Centre of rotation of a maxillary central incisor under orthodontic loading. British Journal of Orthodontics 4: 23–27