DEVELOPMENT OF A 3D FINITE ELEMENT MODEL OF THE MAXILLA FOR NUMERICAL ANALYSIS OF ORTHODONTIC LOAD CASES

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Abstract: Using finite element methods (FEM), a variety of orthodontic problems has been simulated in the past 30 years aiming at the generation of a patient-specific, detailed three-dimensional FE model of teeth and surrounding tissues. The scope of this study was to develop an almost anatomically correct FE model of the maxilla based on a commercial 3D dataset, which can be used for simulation of different orthodontic treatment tasks with only minor expenditure. The surface of the dataset was discretised with tetrahedrons using the FE-programme MSC.Marc®/Mentat®2003r2. Tissue material parameters were taken from previous studies. The model was used to simulate following movements: 1) Simulation of different kinds of movements of blocked incisors to determine the location of the centre of resistance (CR). 2) Contact analyses of molar distalisation employing headgear appliance with molars in different eruption stages. Results: 1) There is no common CR of the incisors. The presented 3D model is suitable to simulate different clinical problems and can be adapted to patient-specific data with minor expenditure. 2) About 30% of the applied force are transferred from the first molar (M1) to the second (M2), if the third molar was not present. The displacements of M1 were twice as large as those of M2.

Introduction

Orthodontic biomechanics deals with the description of the biological reactions of dental structures to the application of orthodontic force systems. Areas of interest in particular are the description of tooth movements using numerical methods, combined experimental and numerical investigations of the behaviour of teeth, the periodontal ligament (PDL) and alveolar bone as well as the analysis and the design of specialised appliances to induce tooth movement.

The treatment of tooth malpositions is achieved by employing specialised orthodontic devices. The force systems are applied to the dental crowns via brackets, inducing bone remodelling processes (see Figures 1 and 2). Translations of teeth of up to 5 mm and rotations of above 10° can be realised. Forces and torques are in the order of 1 N and about 10 Nmm, respectively.

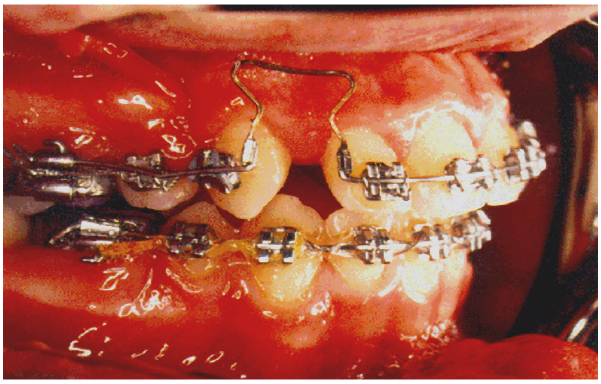

Figure 1: Retraction of the upper incisors. The force system of the appliance must be adjusted such that the rotation of the incisor segment is compensated.

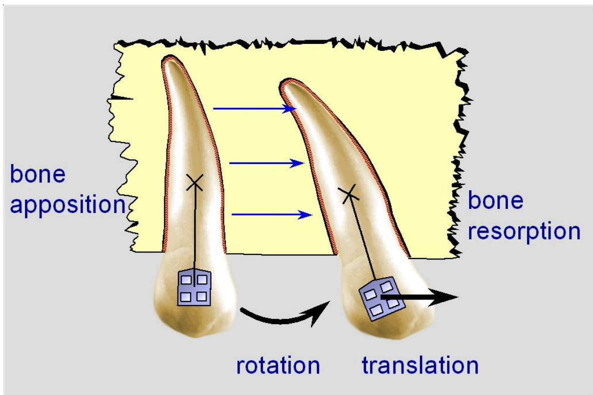

Figure 2: Principle of orthodontic tooth movement.

Physically, the tooth corresponds to a supported, rigid body, that is fixed in the surrounding periodontal ligament. Loading the dental crown with a pure torque, the tooth will execute a pure rotation around a defined point, the so-called centre of resistance (CR). This property was used in a large number of experimental and numerical studies, in order to determine the position of the CR of individual teeth.

Using finite element methods (FEM), a variety of orthodontic load cases have been simulated in the past 30 years [e.g. 1-4]. A major problem in these analyses is the generation of a patient-specific, though detailed three-dimensional FE model of the teeth and the surrounding tissues. Until now, only specific problem-oriented, in part highly idealised FE models have been developed to

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analyse each specific problem. It was the aim of this study, to develop an almost anatomically correct FE model of the human maxilla, which can be used for the simulation of a variety of orthodontic treatment tasks with only minor expenditure. The model as well as two examples are presented.

Materials and Methods

Model generation: Based on the 3D data set of a completely toothed maxilla (Digimation Corp., Louisiana, USA), the FE model of the maxilla with all teeth and surrounding tooth supporting structures (periodontal ligament - PDL and alveolar bone) was generated. The dataset used was a surface model consisting of 4-noded elements. Dental roots and crowns were represented separately, however the alveolus and the PDL were not completely modelled (Figures 3 and 4).

Figure 3: Three dimensional view of the FE model of a maxilla.

Figure 4: Full dental arch with separately displayed crowns and roots.

In a first step, the alveolae were generated by manipulating the alveolar ridge and inserting the scaled surfaces of the given dental roots (Figure 5). The individual teeth were then scaled and reduced in cross section by the thickness of the PDL (0.2 mm) such that they could fit into the alveolar ridge. By doing so for each individual tooth, a periodontal space of correct thickness resulted for all teeth of the dental arch (Figure 5). Subsequently, the individual structures were converted into volume models, and the volumes of teeth, PDL and alveolar bone were combined and connected node by node in the FE programme Marc® /Mentat® 2003r2.

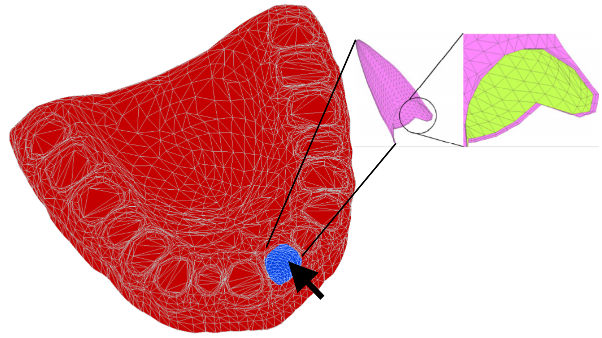


Figure 5: Modelling of an alveolus in the position of an upper right incisor (arrow).

Two orthodontic load cases were simulated. For the first one, idealised brackets were generated on the labial surfaces of the teeth. The four incisors were connected to a block using a steel wire with a retangular cross section of 0.46 x 0.64 mm² (Figure 6). A further simulation was performed with the cross section of 1.38 x 1.92 mm², which clinically cannot be realised. Nevertheless it was used in order to show the effect of blocking the four incisor teeth with wires of varying rigidity. In the second loading case, the influence of different contact forces between the molar teeth was studied. These varying contact forces were simulated to study the influence of the presence of the second molar (M2) and the wisdom teeth (M3) in different eruption stages (Figure 7).

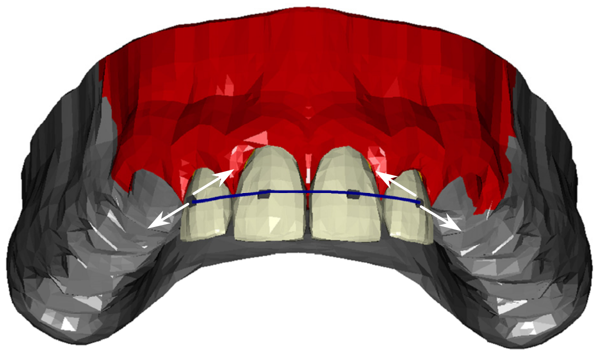


Figure 6: FE model of the maxilla with the four anterior teeth (arrows indicate couples of forces).

The MSC.Marc®/Mentat® software was used for all meshing and calculation purposes. The mesh was generated with the isoparametric 10-noded-tetrahedron element, as nonlinear calculations employing material nonlinearity as well as large displacement and finite rotation analyses were to be carried out. The steel wire in load case 1 was generated with beam elements. Altogether, the maxillary model with all 16 teeth consisted of approximately 124,000 nodes and 655,000 elements. This large number of nodes and elements required a reduction of the number of elements in the simulations of the different load cases, to ensure processing at reasonable time scales. Consequently, only a part of the model was used in the calculations, resulting in models of about 40,000 nodes (e.g. non-grey parts in Figure 6 for case 1).

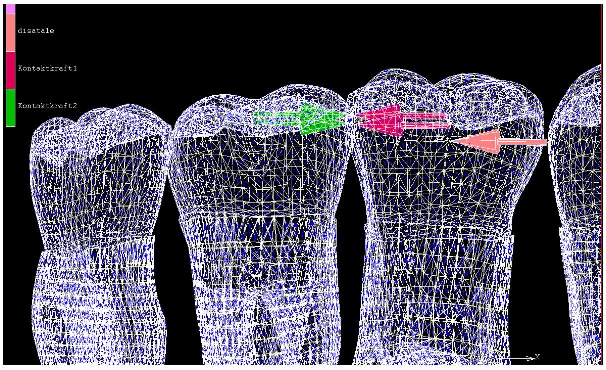


Figure 7: FE model of three molars in fully erupted stage. Contact forces (red and green) and applied orthodontic force (orange) are indicated by arrows.

Calculations with the models: Determination of the centre of resistance of the incisor segment was achieved by loading the lateral incisors with force couples around the bucco-lingual axis of 10 Nmm each. The maxilla was kept stationay by applying boundary conditions to the alveolar bone. For determination of the position of the centre of resistance, displacements of the individual teeth were recorded. Besides this, normal and shear stresses and strains for each node and element were calculated and listed in order to evaluate the biomechanical behaviour of all structures involved.

Crowding of teeth and resulting increased contact forces was simulated by varying approximal forces from 0.0 N to 10.0 N between the first and the second molar. Subsequently, the influence of the eruption stages of M2 and M3 was investigated by performing a serie of contact analyses with and without these molars. A distalising force between 1.5 N and 7.5 N was applied to the first molar (M1), simulating the effect of a so-called headgear or face-bow. The contact analyses were performed with a frictional coefficient of 0.2 and an adapted coulomb method. The normal contact forces between the teeth and the displacements in the model were recorded (Figure 7).

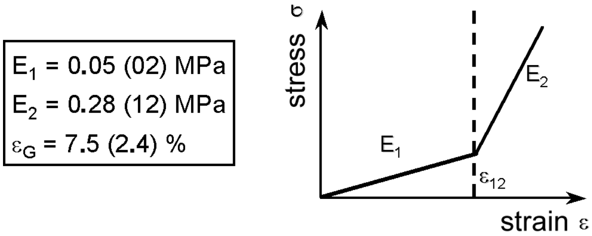
The material parameters of teeth and bone were taken from the literature [5], the non-linear behaviour of the PDL from previous experimental and numerical studies [6, 7]. All parameters are listed in tables 1 and 2.

For all calculations an isotropic and homogenous behaviour was assumed, which of course is an idealisation of the realistic behaviour of the tooth supporting structures. However, this assumption proved to be valid for the orthodontic loading case [4, 7-9].

Table 1: Material parameters of tooth, bone and bracket. Teeth and bone were not sub-divided into enamel / dentin and cortical / spongious bone, respectively.

Material	Young's modulus [MPa]	Poisson's ration μ
Tooth (average value)	20000	0.30
Bone (average value)	2000	0.30
PDL	bilinear	0.30
Bracket (steel)	200000	0.49

Table 2: Parameters describing the non-linear behaiour of the PDL. A bilinear approximation could be determined in previous experimental and numerical investigations.



Results

CR of upper incisors: Figure 8 illustrates the calculated displacements of the four anterior teeth as a result of the applied torques. It is obvious that the crowns of the lateral incisors were displaced clearly further than the central incisors. The maximum displacement of the incisal edges was about 11 µm for the lateral incisors and about 5 µm for the central incisors. Consequently, despite of the archwire blocking the four anterior teeth, they seem to move almost independently.

Figure 9 displays the calculated displacements of the teeth without surrounding PDL and bone in a colour coding. The axes of rotation of the individual teeth can clearly be identified. The upper part of figure 9 depicts the movements in the occlusal and the lower part those in the frontal plane. Assuming that the CR is the point of the dental root with the minimum deflections, each instantaneous axis of rotation can be identified by the blue colour.

As can be seen from figure 9, it is possible to construct a common axis of rotation for all four incisors only in the occlusal plane. In the frontal plane, it seems that the four anterior teeth move entirely independent. Moreover, the axes of rotation of the individual teeth are inclined in the frontal plane. For the individual teeth, the calculations delivered a position of the CR at 9 and 12 mm down the tooth axis and 5 mm distal to the point of load application, i.e. the lateral incisor brackets. Thus,

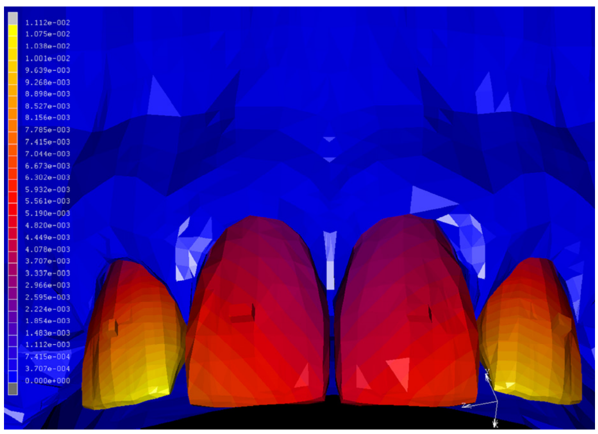


Figure 8: Illustration of the calculated displacements of the maxillary anterior teeth. The yellow areas represent the highest deflections.

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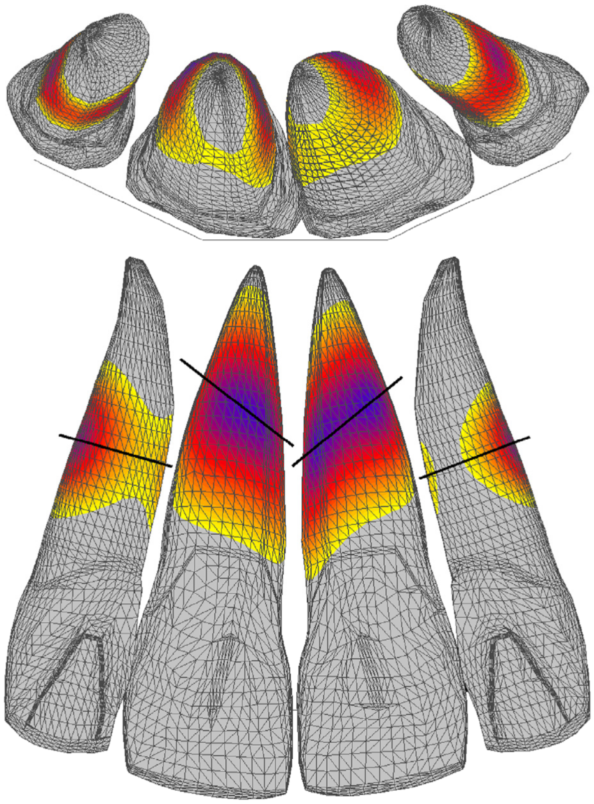

Figure 9: Position of the CR of the anterior segment in occlusal and dorsal view. The axes of rotation were fitted through the areas with minimum shift (blue colour).

there was no common CR for the entire anterior segment. Instead, several isolated CRs for the individual teeth could be determined. These were found to be in a common plane.

An increased rigidity of the blocking arch wire was assumed to reduce the independent movement of the individual teeth. Thus, calculations with an increased cross section of the blocking arch were performed and analysed as before. The effect is shown in Figure 10, in comparison with results from the literature and the results with the regular wire cross section. However, despite the increase of the wire cross section to a threefold value of the largest archwires in clinical use, a perfectly common movement of the four incisors could not be achieved. The individual CRs approached each other, but this shift did not result in a common axis of rotation that was in accordance with the values reported in the literature.

The stress and strain distributions are illustrated in Figure 11. Increased stresses and strains around the roots of the two lateral incisors caused by the increased shift of the teeth can clearly be identified. In particular, due to the highly inhomogeneous load of the tissue around the four anterior teeth, peak stresses and strains occurred at the tips of the roots and at the alveolar crest.

Contact forces between molar teeth: In the initial analysis, the M1 displacements under distalising forces proved to be approximately independent of the contact forces. Thus, the corresponding data is not presented

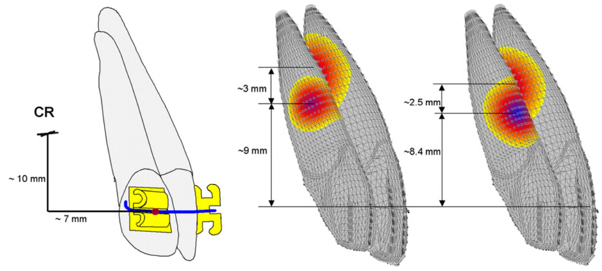

Figure 10: With increasing rigidity of the stabilising arch wire the individual CRs move closer to each other (left: value from literature assuming perfect rigidity, middle: wire cross section 0.40 x 0.56 mm², right: 1.38 x 1.92 mm²).

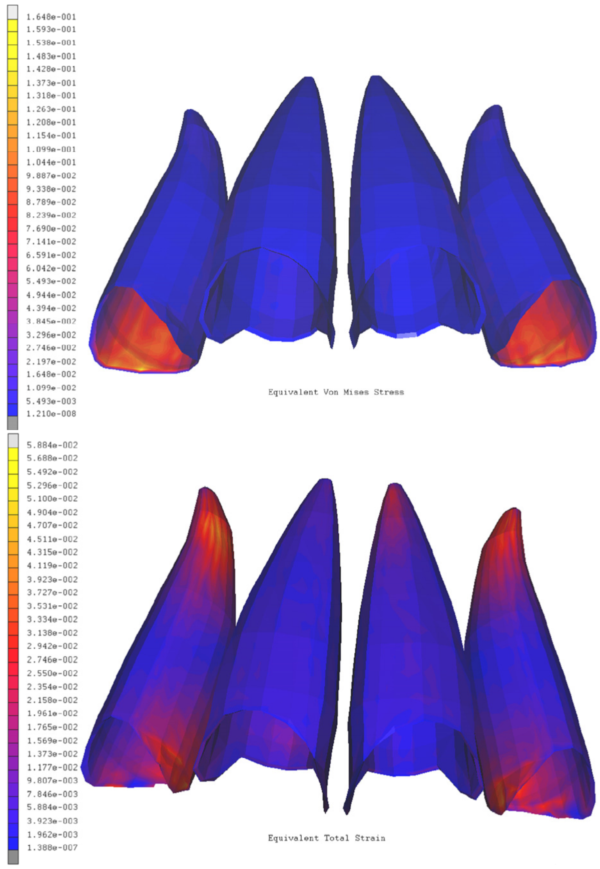

Figure 11: Stress and strain distributions within the PDL surrounding the loaded front teeth. The highest stress values (yellow) are concentrated in the PDL of the lateral incisors.

here. In general, the M3 is clincally not present in cases of headgear treatment. The difference between calculated displacements of M1 and contact forces between M1 and M2 with and without M3 was 6 to 7 %. About 30 % of the force applied to the first molar are transferred to the second molar if the third molar was not present and the calculated displacements of M1 were twice as large as the movement of the M2. A particularly high stress of 1.2 MPa was determined at the alveolar crest in between the molars.

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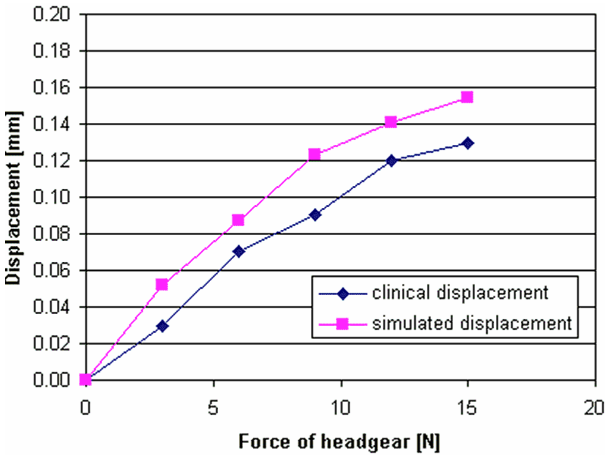

Figure 12: Displacement versus force of headgear of a computation by all molars.

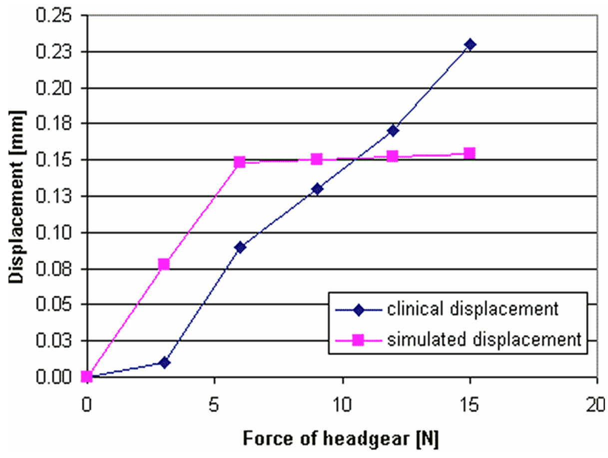

Figure 13: Displacement versus force of headgear of a computation by M1 with extracted M2 and M3.

Figures 12 and 13 display the comparison of the clinical and calculated displacement of the M1 with all molars present (Figure 12) or with extracted M2 and M3 (Figure 13). The simulated displacement against the headgear force with all teeth was about 20 % higher in comparison to the clinical results. When M2 and M3 were not present, the situation was roughly the same at initial loads. However, the M1 showed a slightly higher mobility. In the simulations the M1 displacement was restricted by the contact of the root to the alveolar wall. Thus the displacement nearly stopped at 0.15 mm. Clinically this could not be seen that clear.

Discussion

Using finite element methods, the centre of resistance of the anterior teeth was determined, by analysing the load/deflection characteristic of the upper incisors upon loading with pure torques of 10 Nmm. In contrast to previous studies [e.g. 10], particular emphasis was placed on the common investigation of the four upper incisors in a model of the maxilla with anatomically correct tooth supporting structures and a realistic blocking of the individual teeth with a conventional steel wire.

Previous investigations of the mobility of the incisors and the position of the CR of the anterior segment were based partially on the laws of lever and the assumption of an ideal rigid blocking. However, as the clinical application of an anterior retraction is not focused on the individual teeth, but tends to move the entire segment, the morphology of all four dental roots and the nature of the blocking wire are of particular importance. It may be assumed that the different root morphologies have a crucial influence on the biomechanics of the anterior segment and the simulation of its movement within periodontal space, which was already demonstrated in the calculations of movements of single teeth [7, 11, 12]. This assumption is supported by the presented results. In clinical practice it seems to be impossible to realise an ideal, completely rigid segment of the upper incisors. As the roots of the central incisors have a clearly higher volume than the roots of the lateral incisors, they have a higher resistance towards a displacement within periodontal space. Consequently, this results in torsion of the archwire between the lateral and the central incisors and the central incisors are loaded with smaller force systems than the lateral incisors. This in turn results in increased loading of the lateral incisors. The consequences for clinical application are twofold:

First, the planned tooth movement will not proceed as desired. Concerning the central incisors, the calculated CRs were located 2 - 3 mm further towards the apex than reported in the literature, whereas the CR of the lateral incisors was located at the suggested height (Figure 10). The diverging positions of the CRs of lateral and central incisors in the anterior segment will result in an increased tipping especially of the central incisors. Moreover, it may be assumed that the lateral incisors will initially move faster than the central incisors due to a higher bone remodelling rate, which in turn is based on the assumption that orthodontic bone modelling and remodelling is related to PDL strains [13-16]. The strain in the PDL of the lateral incisors, however, is significantly higher due to the inhomogeneous shift of the individual teeth in the anterior segment (Figure 11). A clinical consequence might be that the lateral incisors will 'take the lead' over the central incisors, within the restriction defined by the elasticity of the stabilising arch wire. The behaviour described above is in good accordance with the observations of previous clinical studies [17, 18].

Second, the larger displacement and higher strains of the lateral incisors harbour the danger that the stresses in the PDL could reach unphysiologically high values. Comparing the calculated maximum stresses in the PDL with the capillary blood pressure (1.2 kPa), it is obvious that the applied force system can result in stresses that are two orders of magnitude higher. Ideally, the stresses in the PDL, resulting from an orthodontic force system, should not markedly exceed the capillary blood pressure as this might cause pain and result in root resorption. Although the tooth supporting apparatus is able to repair minor defects itself, these risks should be kept in mind.

Concerning the simulations of the biomechanical behaviour of the molars, it can be stated that the calculated

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displacements with and without the M3 could be validated by the clinical results. The discrepancy between theoretical and clinical results can be explained by the fact that the maxillary geometry of the FE model is based on adult morphology and the clinical data was taken from adolescents. The break in the theoretical curve of Figure 13 can be explained by a contact of the dental root with the alveolar bone in the region of the furcation due to an increased tipping of M1 without the neighbouring teeth. Clinically, this does not occur, which means that for an imporved simulation of clinical situations the FE model must be adopted to the geometry of a patients dental arch.

Conclusions

The presented study introduced examples for the use of theoretical biomechanical models with direct clinical relevance or implication. Although it must be stated that such theoretical numerical models have severe restrictions with respect to their representation of living biological structures, it seems that the rather large amount of time and processing power resulting from the high complexity of the FE model appears to be justified in the light of the presented results. Further simulations of complex and anatomically correct tooth-jaw models are needed.

As a direct conclusion from the results of these simulations, it can be recommended that the clinician should no longer use the assumption of a common centre of resistance of the upper anterior segment. Planning of anterior retraction should be done with the presented values in mind. The 3D model developed in this study is suitable to simulate different clinical problems and can be adapted to patient-specific data with minor expenditure.

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