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Biomechanical finite-element investigation of the position of the centre of resistance of the upper incisors

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SUMMARY The position of the centre of resistance (CR) is an essential parameter regarding the planning of orthodontic tooth movements. In the present investigation, the combined CR of the upper four incisors was determined numerically using the finite-element (FE) method.

Based on a commercially available three-dimensional data set of a maxilla, including all 16 teeth, as well as known and earlier determined material parameters, FE models of the upper incisors and their surrounding tooth-supporting structures were generated. In the FE system, the model of the anterior segment was loaded with torques of 10 Nmm each at the lateral incisors.

The FE model indicated that the individual incisors moved independently, although they were blocked with a steel wire of dimension $0.46 \times 0.65\ mm^2$. The individual CRs were located at 5 mm distal and 9 and 12 mm apical to the centre of the lateral brackets. Thus, the classical view of a combined CR for the anterior segment was disproved and the planning of orthodontic tooth movements of the upper incisors should no longer be based on that concept.

Introduction

Orthodontic biomechanics describes the biological reactions of the dental structures in the interaction with mechanical forces in orthodontics. Particular emphasis is laid on the description of tooth movements with numerical methods, the combined experimental and numerical analysis of the behaviour of teeth, periodontal ligament (PDL) and surrounding bone, as well as the analysis and design of special treatment elements (Burstone, 1962; Bourauel *et al.*, 1992; Hinterkausen *et al.*, 1998). The location of the centre of resistance (CR) is considered an essential factor in the planning of orthodontic tooth movements. With exact knowledge of the CR, the force system that must be applied to the crown of the tooth to achieve the desired tooth movement can be determined.

Mechanically, a tooth is a supported, rigid body, with its support in the surrounding tissue. If a tooth crown is loaded with a force couple, it will rotate around a well-defined axis, the so-called CR. This mechanical property has been used in a number of experimental and numerical studies to determine the position of the CR of single teeth, in part using highly idealized tooth models (e.g. Burstone and Pryputniewicz, 1980; Dermaut *et al.*, 1986; Vollmer *et al.*, 1999).

The clinician, however, is frequently confronted with the task of moving an entire segment consisting, for example, of the four anterior teeth (Figure 1). Experimental or numerical research concerning the location of the CR of the anterior segment has only been reported in a few studies (e.g. Burstone *et al.*, 1981; Vanden Bulcke *et al.*, 1987; Melsen *et al.*, 1990; Pedersen *et al.*, 1991). Based on these investigations, empirical rules have been derived taking clinical observations also into consideration. Using lever principles, the position of the CR of the four anterior teeth was calculated by superposition of the CR of the individual anterior teeth (Vanden Bulcke *et al.*, 1987; Bauer *et al.*, 1992). This rule of thumb states the position of the CR of the anterior segment to be at 9–10 mm apically and approximately 7 mm distally to the lateral incisor bracket (Gjessing, 1992). However, in the clinical application of this concept, retraction of the anterior segment often results in minor but significant side-effects. In a previous clinical experimental study, anterior retraction was carried out with a precisely calibrated retraction arch (Bourauel and Drescher, 1994). The clinical result indicated that the desired bodily movement of the anterior segment could not be achieved without further correction of the force system, although the CR was assumed to be at the position stated above and an experimental test proved that the force system was appropriate for bodily movement of the segment. Therefore, it can be concluded that the position of the CR of the anterior segment is still uncertain.

The aim of the present study was to investigate the load–deflection behaviour of the upper anterior teeth using finite-element (FE) methods to determine the combined CR of the anterior segment. Particular emphasis was placed on the anatomically correct reconstruction of the geometry of the root morphology and the surrounding PDL, as these have a crucial influence on the reaction of the teeth to the applied force systems (Burstone, 1962).

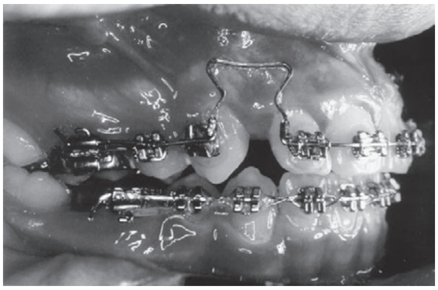

Figure 1 Retraction of the upper incisors using a superelastic spring.

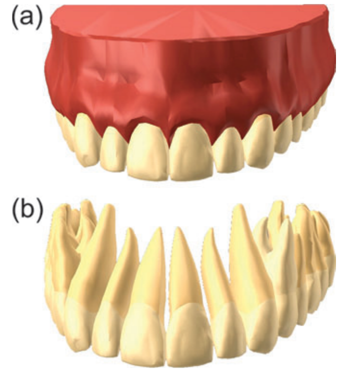

Figure 2 Three-dimensional view of a maxillary finite-element model (a) and all 16 upper teeth (b). The almost anatomically correct geometry of each individual root and crown is represented separately.

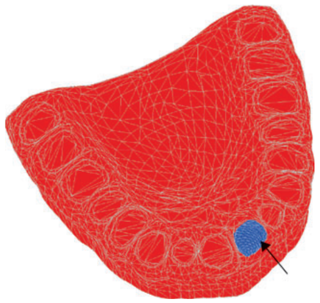

Figure 3 Modelling of an alveolus in the position of an upper right incisor (arrow).

Material and methods

Generation of the models

FE models of the upper central and lateral incisors and the surrounding tooth-supporting structures (PDL and alveolar bone) were generated on the basis of the three-dimensional data set of a maxilla with all 16 teeth (Digimation Corp., St Rose, Louisiana, USA). The data set used was a surface model consisting of four-noded elements. Tooth root and crown were represented separately, but the tooth-supporting structures were not completely modelled (see Figure 2a,b). In a first step, the alveolar process was generated into the surface model of the maxillary bone. This was achieved by fitting the given tooth roots so that anatomically correct alveolar resulted (Figure 3). The individual teeth were then reduced in their cross-section by an average of 0.2 mm so that they could fit into the bone model, and a periodontal space of appropriate thickness resulted. The periodontal space of each individual tooth was created from the surface models of the alveolae and the respective tooth root geometry (compare Figures 3 and 8).

Subsequently, the individual structures were converted into volume models, and alveolar bone, teeth, and the surrounding PDL were integrated and connected node by node in the FE system (Marc/Mentat® 2003, MSC Software Corp., Santa Ana, California, USA). In the next step, idealized brackets for load application were generated on the labial surfaces of the teeth. Blocking of the four incisors was realized by inserting a steel wire with a rectangular cross-section of 0.46×0.64 mm^2 (Figure 4a). In a further FE model, the wire cross-section was tripled to 1.38×1.92 mm^2.

The Marc/Mentat® software was used for all meshing and calculation purposes. The mesh was generated with the isoparametric 10-noded tetrahedral element because non-linear calculations with material non-linearity as well as large displacements and finite rotations were to be carried out. The steel wire was generated with beam elements. Altogether, the maxillary model with its four teeth was made up from approximately 75 000 nodes and 366 000 elements. This large number of nodes and elements required a reduction of the number of elements to ensure processing in a reasonable time scale. Thus, only a part of the bone structure was used in the calculations (Figure 4), resulting in a model of about 150 000 elements.

Calculations with the models

To determine the CR, the anterior teeth were loaded with force couples around the bucco-lingual axis of 10 Nmm each via lateral incisor brackets, while boundary conditions were applied to the bone to keep it stationary (Figure 4b). The material parameters of teeth and bone were taken from the literature (Abé *et al.*, 1996), the non-linear behaviour of the PDL from previous experimental and numerical studies (Vollmer *et al.*, 2000; Poppe *et al.*, 2002). The parameters are listed in Tables 1 and 2. For all calculations, an isotropic and homogenous behaviour was assumed, which, of course, is an idealization of the realistic behaviour of the tooth-supporting structures. Especially, isotropic and symmetric behaviour in compression and tension was assumed for the PDL. Of course, this does not reflect perfectly the complex structure and behaviour of the PDL. However, in combined experimental and numerical studies, this assumption proved to be valid for orthodontic loading and was sufficient to describe initial tooth displacements (Vollmer *et al.*, 1999; Poppe *et al.*, 2002; Kawarizadeh *et al.*, 2004).

For determination of the position of the CR, 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.

Results

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

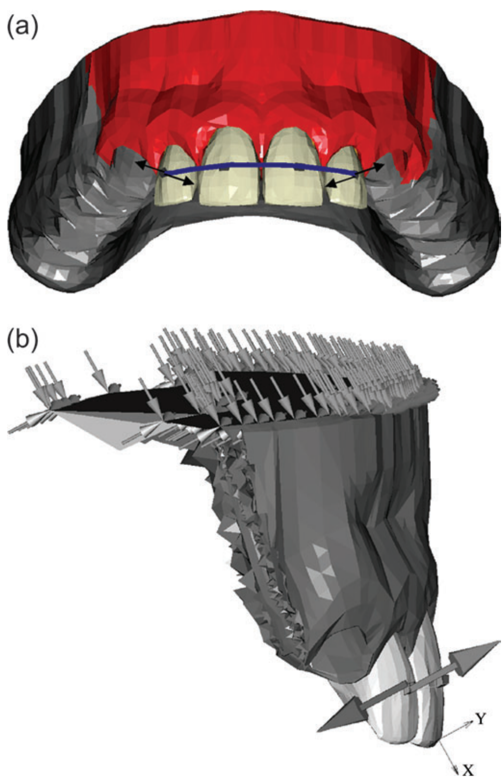

Figure 4 (a) Finite-element (FE) model of the maxilla with the four anterior teeth. Arrows indicate the force couples. Only the anterior part of the model was used for the calculations. (b) Orthogonal view of the anterior part of the FE model. The force couples are indicated by the arrows at the lateral incisor bracket. Boundary conditions are indicated by the arrows on top of the model. All three translational degrees of freedom of the nodes on the outer surface of the bone are fixed.

Table 1 Material parameters of tooth, bone, and bracket. Teeth and bone were not differentiated into enamel/dentine and cortical/spongious bone.

Material	Young's modulus (MPa)	Poisson's ration μ
Tooth (average value)	20 000	0.30
Bone (average value)	2000	0.30
Periodontal ligament	bilinear	0.30
Bracket (steel)	200 000	0.30

The calculated movements of the teeth without the surrounding PDL and bone using colour coding are displayed in Figure 6. Thus, the axes of rotations of the individual teeth can clearly be identified. Figure 6a depicts the movements in the occlusal and Figure 6b those in the frontal plane. Assuming that the CR is the point of the tooth root with the minimal deflections, each instantaneous axis of rotation can be identified by the blue colour.

As can be seen from Figure 6, 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 independently. 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 apical and 5 mm distal to the point of load application, i.e. the lateral incisor brackets. Thus, for the entire anterior segment, there were several isolated CRs instead of a single, common CR. However, these were found in a common plane.

It was considered that an increase in rigidity of the blocking archwire might reduce the independent movement of the individual teeth (Figure 7). However, despite the increase of the wire cross-section to a threefold value of the largest archwires in clinical use, a completely common movement of the four incisors could not be achieved. The individual CRs approached each other, but this shift did not

Table 2 Parameters for the description of the non-linear behaviour of the periodontal ligament. The behaviour of the tooth-supporting apparatus could be described with this bilinear approximation in experimental and numerical investigations.

Material parameter	Value
E_1 (MPa)	0.05
E_2 (MPa)	0.2
ε_{12} (%)	7.5

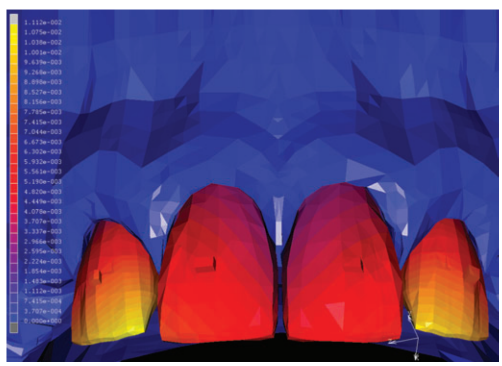

Figure 5 Illustration of the calculated displacements of the maxillary anterior teeth. The yellow areas represent the highest deflections.

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 8. Increased stresses and strains around the roots of the two lateral teeth caused by the increased shift of the lateral incisors can be clearly identified. In particular, the highest values of strains occurred at the tips of the roots and at the alveolar crest.

Discussion

In the present study, the load–deflection characteristic of the upper anterior teeth on loading with pure torques of 10 Nmm was investigated using FE methods. In contrast to previous theoretical studies (Geramy, 2000), particular emphasis was placed on the common investigation of the four teeth 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.

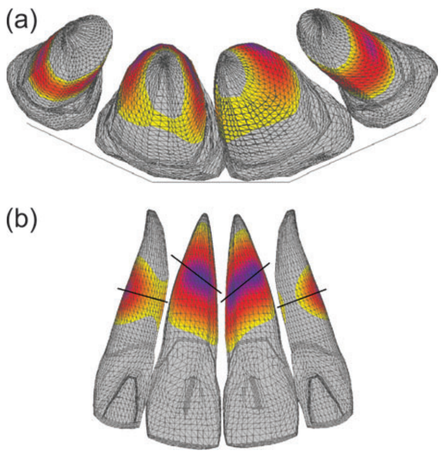

Figure 6 Position of the centre of resistance of the anterior segment in occlusal and dorsal view. The axes of rotation were fitted through the areas with minimal shift (presented in blue).

Previous theoretical, numerical, or clinical investigations of the mobility of the anterior teeth and the position of the CR of the anterior segment consisting of two-, four-, or six-tooth units were based on the laws of lever and the assumption of an ideal rigid blocking of the individual teeth (Vanden Bulcke *et al.*, 1987; Melsen *et al.*, 1990; Pedersen *et al.*, 1991; Yoshida *et al.*, 2001). Based on the methods used, the studies delivered partly inconsistent results, especially with regard to the CR of the four- and six-tooth segments. The most relevant aspect in the cited studies was the definition of a common CR for the segments, having a position between 8.1 and 14.7 mm apically to the incisal edge (Vanden Bulcke *et al.*, 1987; Melsen *et al.*, 1990; Pedersen *et al.*, 1991). In contrast, in the present study, no common CR for the anterior segment was found. The CRs for the lateral incisors were located 13 mm apically from the point of force application and for the central incisors 16 mm apically. For the lateral incisor, the bracket was approximately 2 mm apical to the incisal edge. This means that the position of the CRs of the lateral incisors is comparable with the CR of the anterior segment reported by Yoshida *et al.* (2001), whereas the central incisors have a CR that is located more than 3 mm further apical. Clinically, it might be expected that an increased tipping occurs, which must be counteracted by an uprighting moment.

As the clinical application of anterior retraction is not focused on individual teeth, but instead attempts to move the entire segment, the morphology of all four tooth roots and the nature of the blocking wire are of particular importance. It may thus 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 the periodontal space, which has already been demonstrated in the calculations of movements of single teeth (Tanne *et al.*, 1991; Poppe *et al.*, 2002; Turk *et al.*, 2005). This assumption is supported by the present results.

In clinical practice, it seems to be impossible to realize an ideal, completely rigid segment of the anterior teeth.

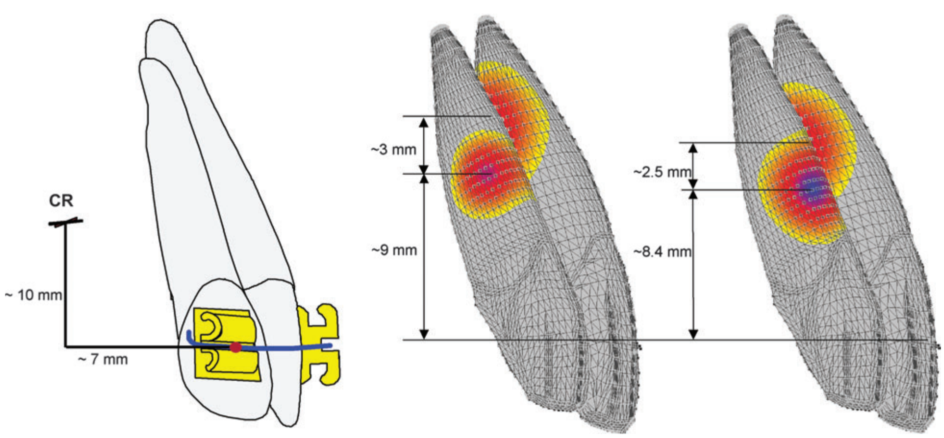

Figure 7 With increasing rigidity of the stabilizing archwire (right side: wire cross-section 1.38 × 1.92 mm^2), the individual centres of resistance (CRs) move closer to each other.

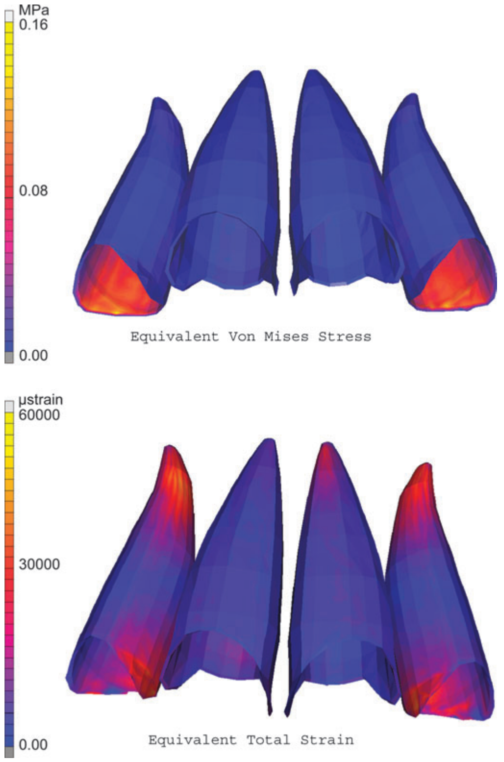

Figure 8 Representation of the periodontal ligament (PDL) with stress and strain distributions surrounding the loaded anterior teeth. The highest stress values (blue) are concentrated in the PDL of the lateral incisors.

Furthermore, in clinical cases, archwires have a certain play in the bracket slots (e.g. Meling *et al.*, 1998). As the simulations did not take any wire/bracket play into account, the results may represent the theoretical best-case scenario that cannot be achieved clinically. Moreover, 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 displacement within the periodontal space. This fact, together with the restricted rigidity and the wire/bracket play, results in torsion of the archwire between the lateral and the central incisors and a torque loss especially at the central incisors. Thus, the central incisors are loaded with smaller force systems than the lateral incisors. This in turn means that the lateral incisors are loaded higher by the applied force system than planned by the clinician.

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 stabilizing archwire. The behaviour described above is in good accordance with the observations of previous clinical studies (Bauer *et al.*, 1992; Bourauel and Drescher, 1994). Tooth movements performed in those studies were planned using the 'classical' value for the position of the CR of the anterior teeth. The clinical result was a remarkable tipping and an intrusion of the anterior segment, although the forces and moments were correctly adjusted to achieve a pure retraction of the anterior teeth. This is also frequently reported by clinicians.

Regarding the material parameters of the PDL, it must be stated that values reported in the literature differ significantly by a factor of up to 10^4 (for review, see e.g. Ziegler *et al.*, 2005). These differences have their reason in the experimental designs, a large variation in the complexity and geometry of the numerical models, and different assumptions regarding the mechanical modelling of the PDL in theoretical investigations. However, as it was not within the scope of the present study to exactly model the inner structure of the PDL but instead to describe its overall behaviour, it was assumed that this idealization is sufficient to simulate initial tooth displacements theoretically. This, in particular, is in accordance with the work described by Cattaneo *et al.* (2003). In addition to the non-linear elasticity of the PDL, those authors used an asymmetric behaviour in compression and tension. They found a strong dependence of bone stresses in pressure and tension regions due to the asymmetric behaviour. However, the influence of asymmetry and anisotropy on tooth displacement seems to be of minor importance, as the calculated deflections of the crown were of a similar magnitude as in previous studies (Cattaneo *et al.*, 2003). Thus, it can be concluded that although this theoretical model is limited by a number of idealizations, the calculated individual CRs of the anterior teeth are of direct clinical impact and should be considered in treatment planning rather than the classical concept of a common CR.

Conclusions

The present study introduces an interesting example for the use of theoretical biomechanical models with direct clinical relevance or implication. Although it must be stated that such theoretical numerical models have 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 appear justified in the light of the results. Further simulations of complex and anatomically correct tooth–jaw models are needed. As a direct consequence of the results of this simulation, it can be recommended that the clinician should no longer use the assumption of a common CR of the upper anterior segment. Planning of anterior retraction should be undertaken bearing in mind the new presented values.

Clinically, the consequences are as follows:

1. The diverging positions of the CRs of the lateral and central incisors in the anterior segment will result in increased tipping, especially of the central teeth. This must be compensated by an additional uprighting moment.
2. The strain in the PDL of the lateral incisors is significantly higher than at the central incisors due to the inhomogeneous shift of the individual teeth in the anterior segment. Thus,

it may be assumed that the lateral incisors might initially move faster than the central incisors due to a higher bone remodelling rate, which in turn is based on the assumption that bone modelling and remodelling in orthodontics is related to PDL strains (Katona *et al.*, 1995; Middleton *et al.*, 1996; Bourauel *et al.*, 2000; Kawarizadeh *et al.*, 2003; Toms and Eberhardt, 2003).
3. The larger displacements and higher strains around the lateral incisors increases the risk that the stresses in the PDL could reach unphysiologically high values.
4. Comparing calculated maximum stresses in the PDL with capillary blood pressure (0.0012 MPa), it is obvious that the applied force system can result in stresses that are two orders of magnitude above this value.

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