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EFFECTS OF BONE REMODELLING DURING TOOTH MOVEMENT

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Abstract: For a long time bone remodelling theories have been applied to various bones of the human skeleton with great success. Other authors, such as Middelton, Jones & Wilson [6], Provatidis [5], or Hempowitz [9] showed that the previous bone remodelling theories are not directly applicable to the orthodontic tooth movement due to the complex anatomy of the periodontium and the large changes of geometry. This paper deals with special problems of bone remodelling concerning the mandible and its transformation with finite element method (FEM). With the help of a two-dimensional tooth model different effects of bone remodelling were studied and, if possible, compared to clinical results. The effects of the remodelling function during tooth movement and the different processes of tooth movement are discussed.

Key words: initial tooth movement, periodontal ligament, finite element method, material properties, *in vivo* measurements

Introduction

The ability of bone to adapt to the respective loading situation by means of remodelling processes is exploited in orthodontics to upright teeth or to move them to other positions. The basic processes are interactions between mechanical stimuli acting on the cells and the induced biochemical cell functions. In 1892 Wolff [1, 2] developed the first mathematical formulation of these facts. He postulated that each change in an internal structure or external shape of a bone is effected in accordance with mathematical laws. Several teams use an internal bone remodelling function describing the bone density in dependence on different factors and correlate it with the Young’s modulus. There, the strain energy density [3, 4] is considered as biological stimulus. This is mostly applied to calculations with femoral bones because their geometry do not change throught the remodelling. In the case of the jawbone the internal remodelling cannot be realized in an easy way because of the complex anatomy of the periodontium and the large changes of geometry during the remodelling process. In the case of the external remodelling, bone remodelling is effected by adapting the surface of the alveolus to the respective loading situation. The response of bone tissue to external loading in the jaw is inverse to other bones. In loaded areas jawbone destruction occurs, whereas a femoral bone under load generates bonemass and gets a higher density. Consequently, the bone remodelling functions applied for the jaw must be changed. For the bone remodelling signal scalars such as the strain energy density are applied to the femoral bone. This cannot be transformed to jawbone remodelling functions, because a three-dimensional movement takes place which requires sufficient information on the direction. Provatidis [5] shows in analytical calculations that bone remodelling perpendicular to the alveolar surface unnaturally deforms the alveolus. A sufficient information on direction can only be obtained by viewing the biomechanical stimuli componentwise. The result of this is a three-dimensional bone remodelling vector that controls tooth movement. The transmission of stimuli is not effected directly from the tooth to the jawbone, but influenced considerably by the characteristics of an intermediate layer - the periodontal ligament (PDL). According to Middelton, Jones & Wilson [6] bone remodelling is rather controlled by the PDL than by the jawbone itself, because there

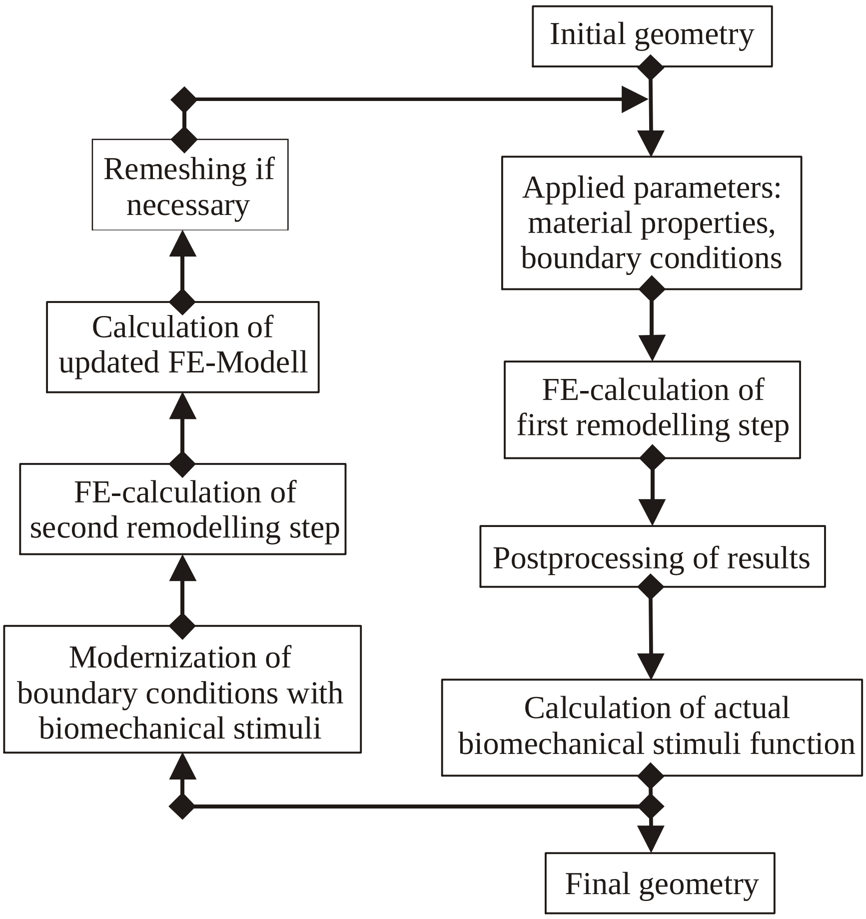

Fig. 1. Representation of the remodelling algorithm.

are considerably larger strains in the PDL than in the jawbone. The result of this is that the biomechanical stimulus is analysed at nodes at the border of alveolar bone to PDL as a mean value. During an orthodontic treatment teeth can move several millimeter. During FEM-simulation this large movement requires special demands on the model and the affiliated mesh. An iterative remodelling algorithm and an adaptive FE-tooth model guarantee high quality and flexibility of this FE-simulation. With a two-dimensional FEM-model the aforementioned phenomena are discussed and parameter studies of different load cases were carried out. In this study three different load cases are presented that clearly show the behaviour of this model. This paper is a component of an DFG (Deutsche Forschungsgemeinschaft) supported project to find the best force system by orthodontic tooth movement through numerical simulations.

External Remodelling Law

Applying a force system to a tooth induces an initial movement. Over a longer time this new tooth position affects a reaction in its embedding and bone remodelling occurs. This initial deflection only considers the first reaction of the periodontium to this force system. An equilibrium between external and internal loadings arises. After some time the tooth moves to a new position through bone remodelling. This new position corresponds to a new equilibrium. According to several investigations, these movements are controlled by strains in the bone [10, 11]. According to Middelton, Jones & Wilson [6] strains in the PDL are responsible for bone remodelling, since only slight strains occur in the jawbone. In the external bone remodelling algorithm applied here, this process was divided into individual

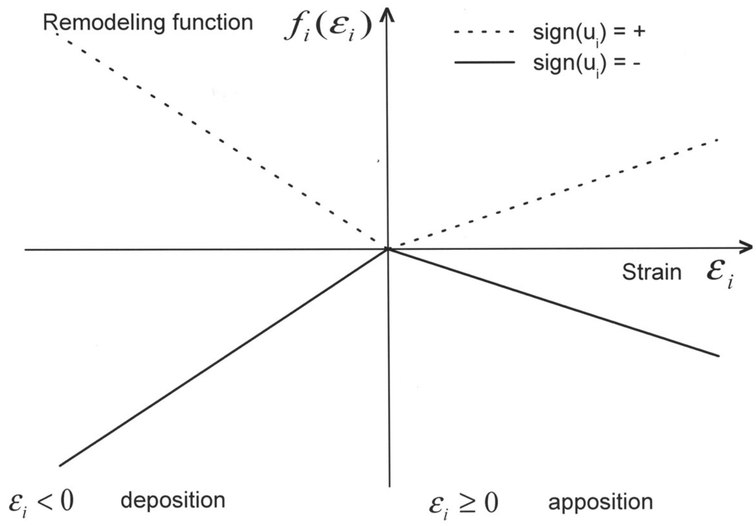

Fig. 2. Graph of a remodelling component in dependence on strain with *i* = *x*, *y*.

independent equilibrium iterations (Fig. 1). The first remodelling step of the initial tooth movement is calculated with FEM. For all nodes situated at the alveolar margin the stimuli function is calculated componentwise from the obtained initial local strain field $\boldsymbol{\varepsilon}$ in the PDL using the following equation ($i = x, y$):

$$f_i(\varepsilon_i, u_i) = \begin{cases} sign(u_i) \cdot |\varepsilon_i| \cdot \kappa_a, & \varepsilon_i \geq 0, \\ sign(u_i) \cdot |\varepsilon_i| \cdot \kappa_d, & \varepsilon_i < 0. \end{cases} \tag{1}$$

The strain component ε_i determines the size of the respective remodelling component f_i. Bone apposition occurs in the extension $\varepsilon_i \geq 0$, κ_a is applied as scaling factor. Compression means bone depositoin, that is scaled with κ_d. The same scaling factors apply to all components. Therefore, the remodelling function is isotropic. The direction of the remodelling vector is determined by the sign of the initial displacement component u_i. Fig. 2 represents a component of the remodelling function. Bone remodelling is controlled by cellular mechanisms and is an active process that requires a certain amount of work. In the second remodelling step the calculated stimuli quantities f_i are applied as componentwise forces to the nodes of the FEM-mesh, at the alveolar margin. In this second step the first initial force system is deleted. By means of these new load factors work is done at the system, and the altered structure is the new initial geometry. This process is repeated iteratively until the desired tooth movement is achieved. The following simulations were aborted at 1500 iterations because the movement is longer than the distance of one tooth-width and it is enough to see the effects. With the help of the applied remodelling function bone remodelling can be controlled separately in the destruction and formation areas by scaling factors κ_d and κ_a. With values of $\kappa_d = 5e^{-3}$ and $\kappa_a = 10e^{-3}$ the numerical simulation becomes stable and the remodelling speed is acceptable. They were constant through all solutions. This way, it is possible to control the deformation of the alveolus and the behaviour of the PDL. It is feasible that the PDL enlarge or scale down through the bone remodelling. Also the shape of the alveolus may be changed during remodelling. To minimize and control these side effects an algorithm changing the PDL stiffness during the second remodelling step is applied.

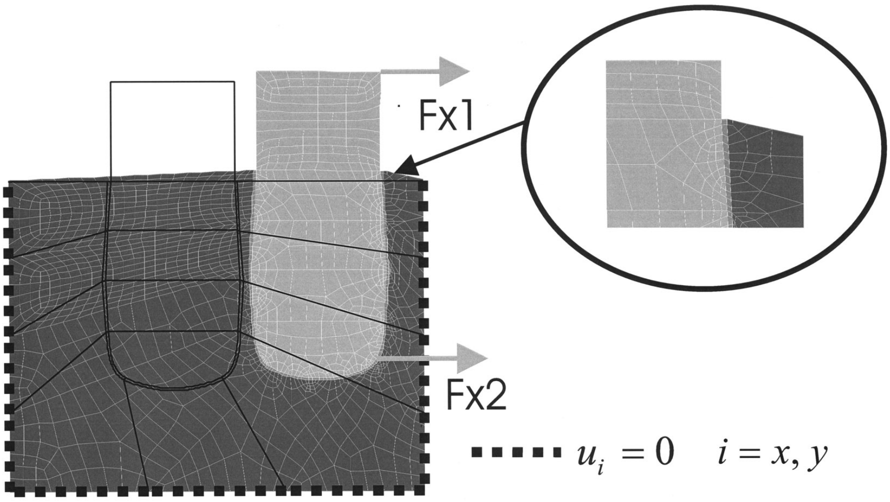

Fig. 3. Starting geometry with points and lines superimposing the FE model changed after 1500 iterations. The enlarged detail shows the periodontal ligament. The boundary conditions and loadings are marked.

FEM-calculation model

Three-dimensional tooth models are necessary for a reliable prediction of tooth movement induced by an external force. Here, a two-dimensional tooth model is used, because the different bone remodelling functions and material parameters can be tested and further developed considerably faster with this model. Later it is planned to pass over to a three-dimensional model with verification through long-term measurements of tooth movement. The model was created and calculated using the Ansys 5.5 FEM software code. The used two-dimensional finite elements have no rotational degrees of freedom. This is the reason why we use two forces as orthodontic force system and do not use customary moments and forces. The lingual lever arm technique [12] implements this orthodontic force system in practice. The evaluated model represents the typical dimensions of an anterior tooth. Special attention was paid to the relation crown height 8 mm to root height 16 mm, because this determines the embedding of the tooth and is therefore a decisive factor for the behaviour of the tooth. The periodontal ligament in the model showed a thickness of 0.25 mm. The large changes in geometry during bone remodelling in the jaw require an adaptive model that allows a new mesh after a number of steps. This was achieved by means of a bone remodelling algorithm that divides the complete tooth movement into several iterations. After a certain number of iterations it is possible to get a new geometry out of the FE-mesh. Then this adapted geometry model has to be covered with a new FE-mesh. Only with this method large tooth movements can be realised with a constant mesh quality during the entire calculation. This ensures that the local information used to calculate the stimuli function is of constant quality. The boundary conditions are displacements and loads, see Fig. 3, here the marked areas have zero displacements during all the simulation. The forces in the first remodelling step are constant, in the second remodelling step they are calculated through the bone remodelling algorithm. Three different load cases were calculated and later presented in the analysis. The different force vectors are also shown in Fig. 3 and are varied in sum.

Material laws

The material properties of the involved tissues (Tab. 1) and types of bone vary considerably in literature. This is the reason why a mean value is used [9]. In this study all materials were treated linear-elastic. It was not distinguished between cortical and spongy bone because of the complexity of the FE-model. Also the assumption of an linear-elastic

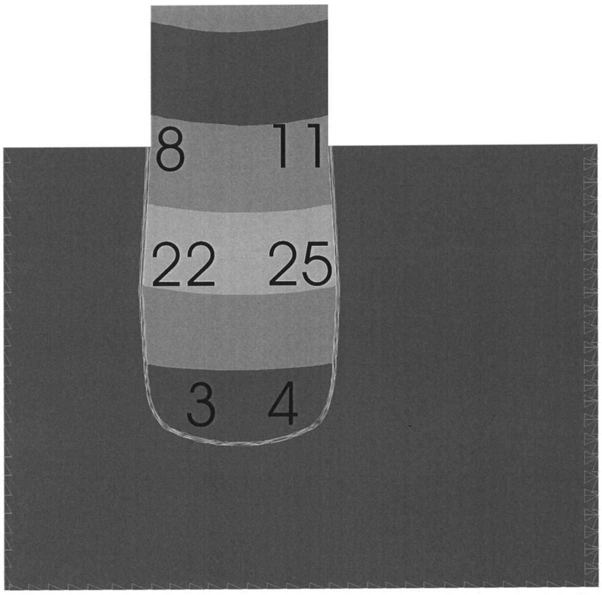

Fig. T1. Starting geometry and initial displacement sum. The numbers correspond to the evaluated nodes.

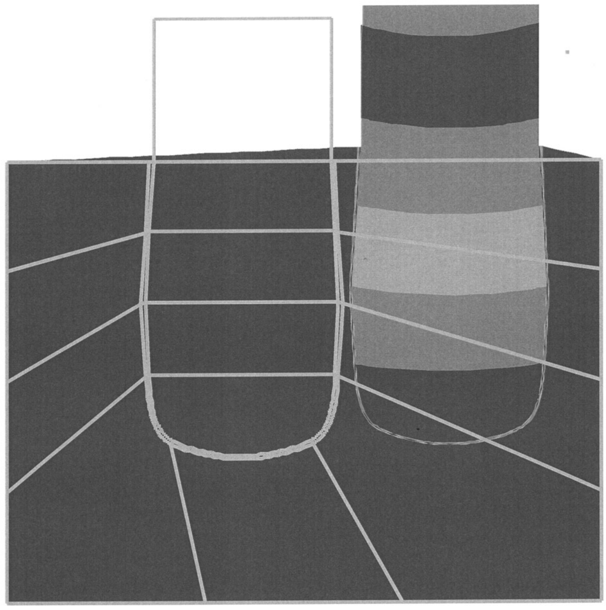

Fig. T2. Geometry and initial displacement sum after 1500 remodelling iterations.

Table 1. Material parameters of the involved tissues.

	Young`s modulus	Poisson`s ratio
Tooth	18.6 GPa	0.35
Jawbone	1 GPa	0.35
PDL	1 MPa	0.45

PDL is a simplification because of the complex anatomy, for further details see for instance Nyashin, Osipov, Bolotova, Nyashin & Simanovskaya [22] or Hempowitz [9].

Results

Three different cases of tooth movement are presented and investigated concerning their clinical correspondence. At first, the case of a pure translation is shown, then a predominant tipping with the effect of a pocket formation, and at last, a translation with a significant tipping. In the following diagrams the local strains and the absolute remodelling movements of defined nodal points are shown. The nodes are viewed at the gingival margin, in the middle and at the lower margin. Fig. T1 shows the numbering. The analysed nodes adjoin to the PDL and the alveolus. The local strains in the PDL were taken into consideration for the remodelling function.

Case of translation: Fx1 = 1 N, Fx2 = 2.7 N

The tooth was loaded by two different forces in the first remodelling step (compare Fig. 3). The force Fx1 is 1 N and the antirotational force Fx2 is 2.7 N. With this force relation, the best possible, pure translation was achieved. It was founded by varying the antirotational force Fx2. Fig. T1 shows the initially loaded nodes with the starting geometry. The centre of rotation is situated far out above the tooth. However, with the FEM it cannot be evaluated

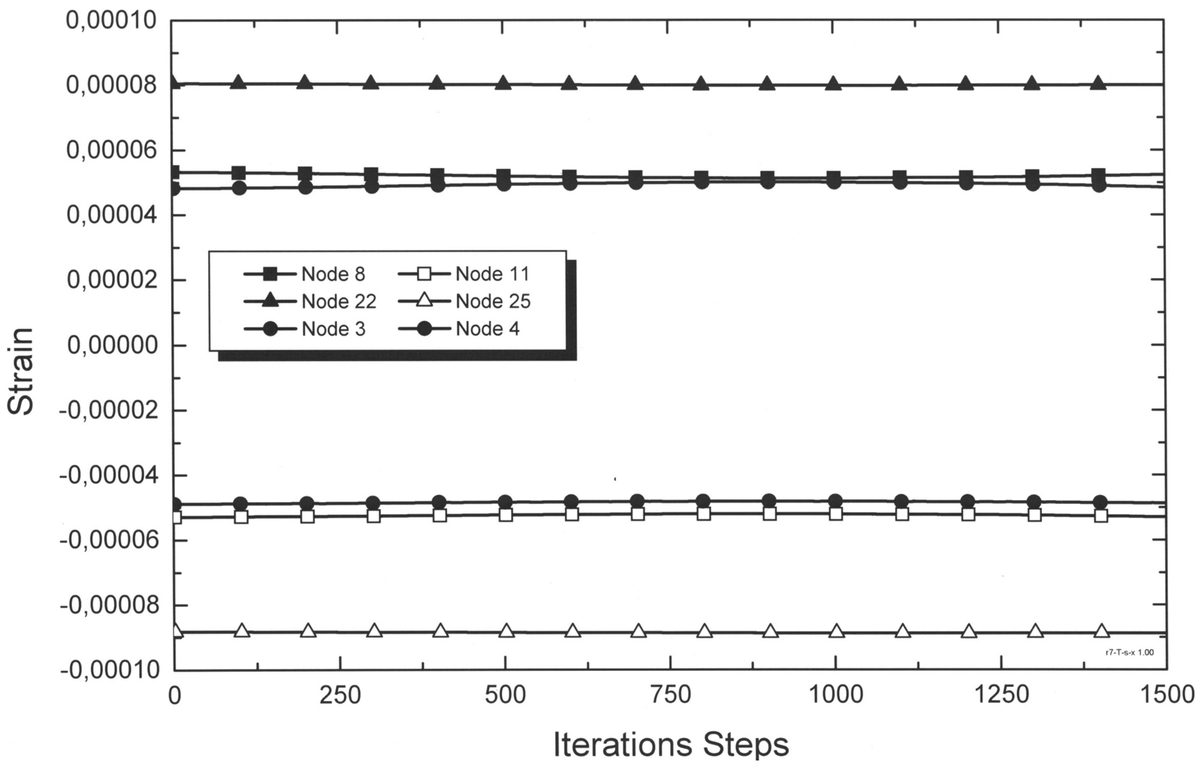

Fig. T3. Course of strain in X-direction over the iteration steps.

where the exact position is. According to Hempowitz [9] and Burstone & Pryputniewicz [19] it is in the infinity. The force is applied directly to the centre of resistance or with suitable combinations of forces and moments to any point on the crown of the tooth. With a relationship of 1/2.7 of upper and lower force as selected here, the centre of resistance must be located in the lower half of the root of the tooth. It is not possible to solve the exact position with this model. Osipenko, Nyashin & Nyashin [7] proposed a method to find center of resistance and center of rotation. Even after 1500 remodelling iterations and a movement in X-direction over the distance of one tooth (Fig. T2), the position of the centre of rotation is still unchanged far out of the tooth. A rotation of -0.5 degree around the Z-axis occurred during this movement and this almost corresponds to a pure translation. The strain component in X-direction (Fig. T3) remains constant during the entire tooth movement, and as expected, the movement of the nodes (Fig. T5) shows an almost linear course. The Y-strain component (Fig. T4) has a linear course. Only the strain at the nodes 3 and 4 indicate a Y-component in remodelling. In the case of an ideal force application in the centre of resistance an insignificant Y-strain component would be expected. The consequence of this would be a pure translation. The absolute movement in Y-direction (Fig. T6) is by the factor 12 lower than the X-component. The regular movement of all nodes in Y-direction is effected by the coupling in the second remodelling step. In this way, an excessive deformation of the alveolus can be avoided. The thickness of the PDL is not influenced either. Consequently during treatment a slight intrusive force magnitude must be applied in order to avoid an extrusion of the tooth. This is the effect of a centre of resistance that is never known exactly. In this way, an ideal force/moment combination cannot be found. There always occurs an extrusive component during tooth movement.

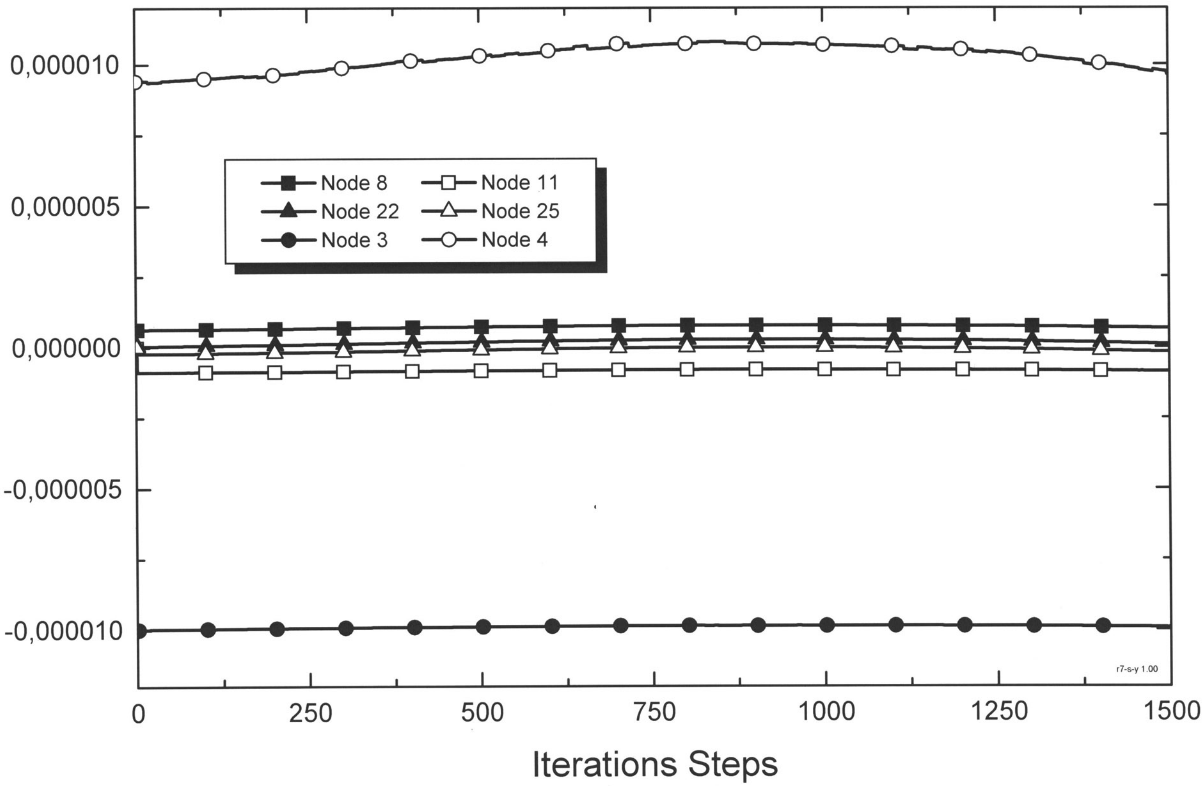

Fig. T4. Course of strain in Y-direction over the iteration steps.

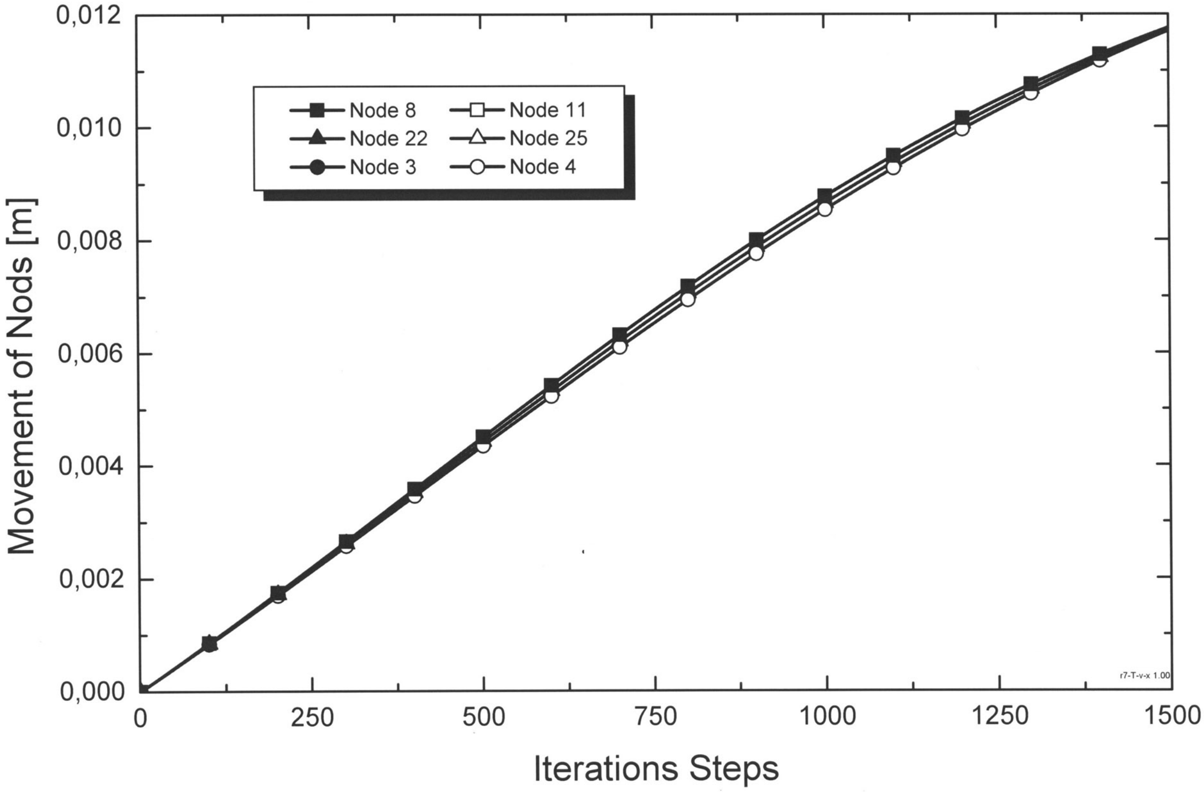

Fig. T5. Course of movement in X-direction over the iteration steps.

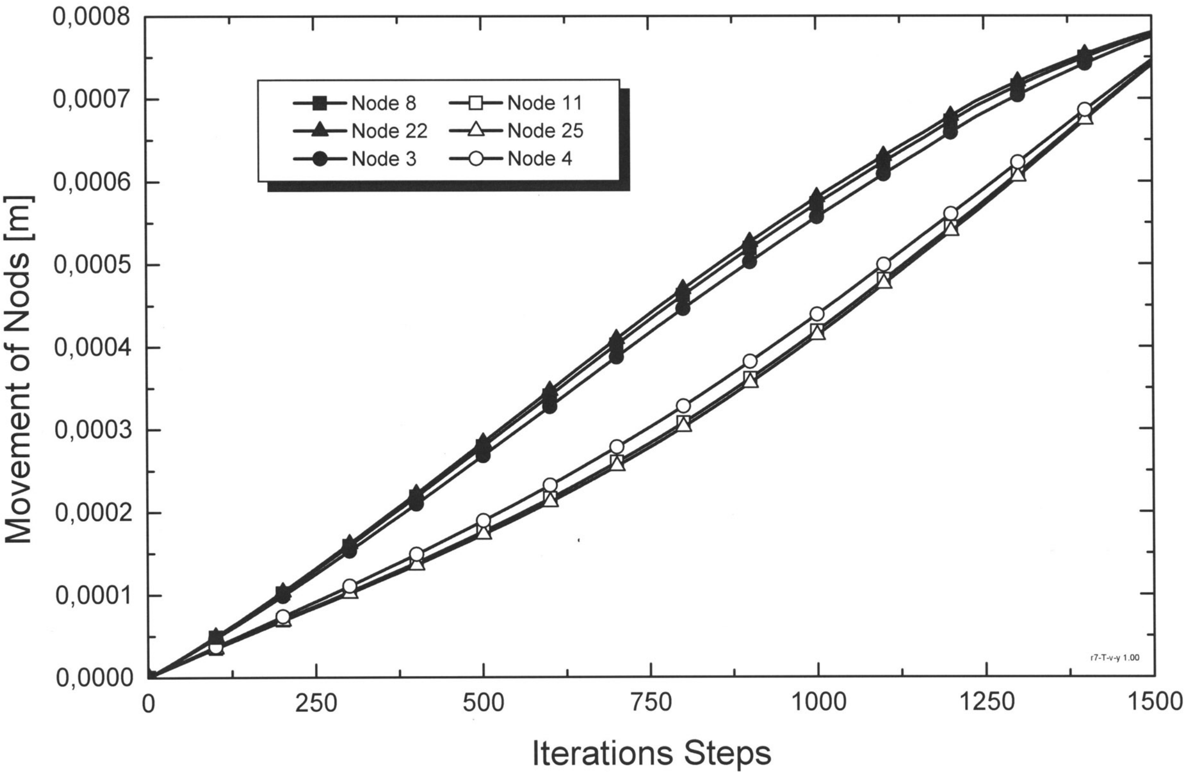

Fig. T6. Course of movement in Y-direction over the iteration steps.

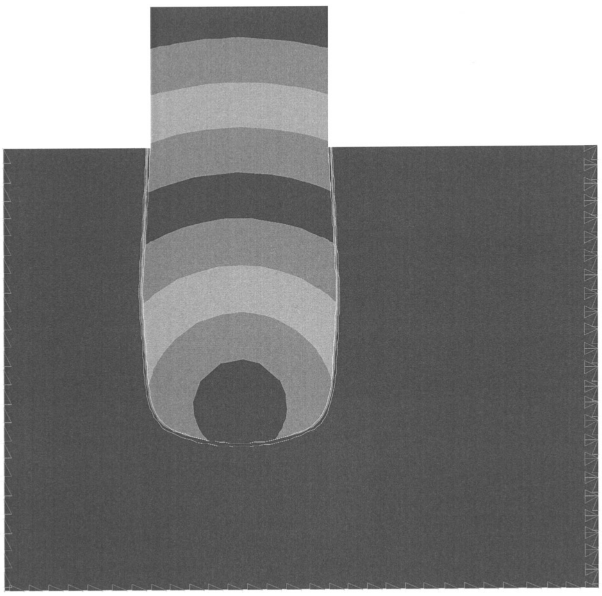

Fig. K1. Starting geometry with initial displacement sum.

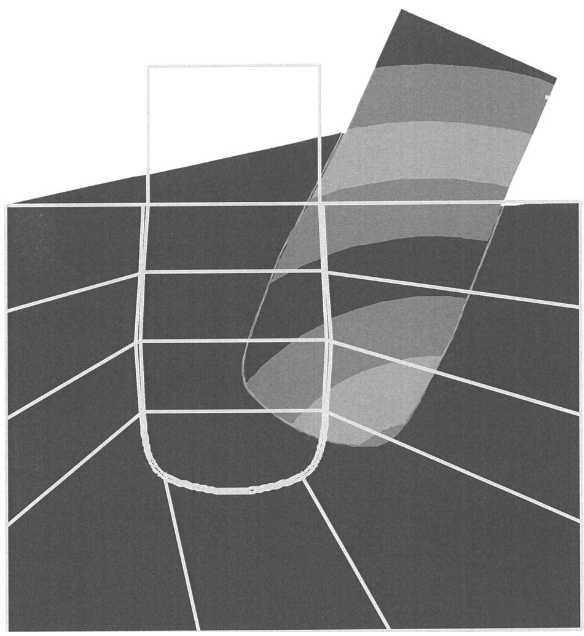

Fig. K2: Geometry and initial displacement sum after 1500 remodelling iterations.

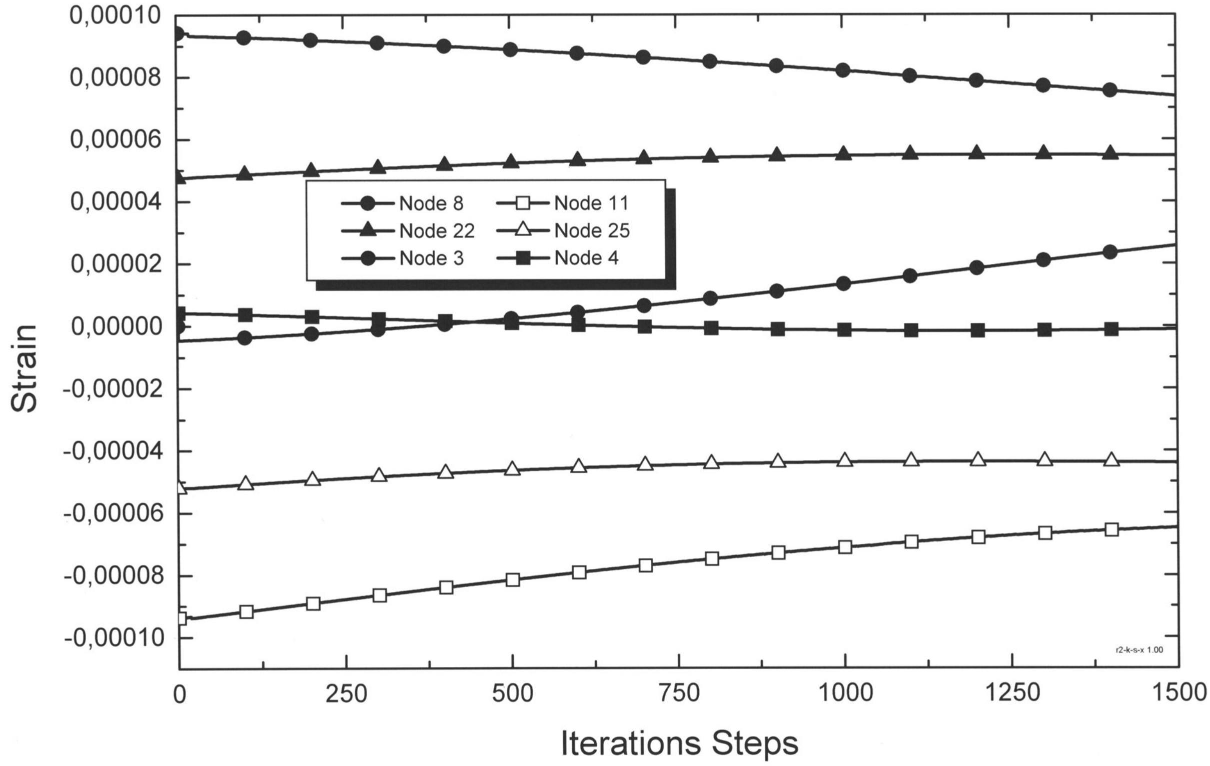

Fig. K3. Course of strain in X-direction over the iteration steps.

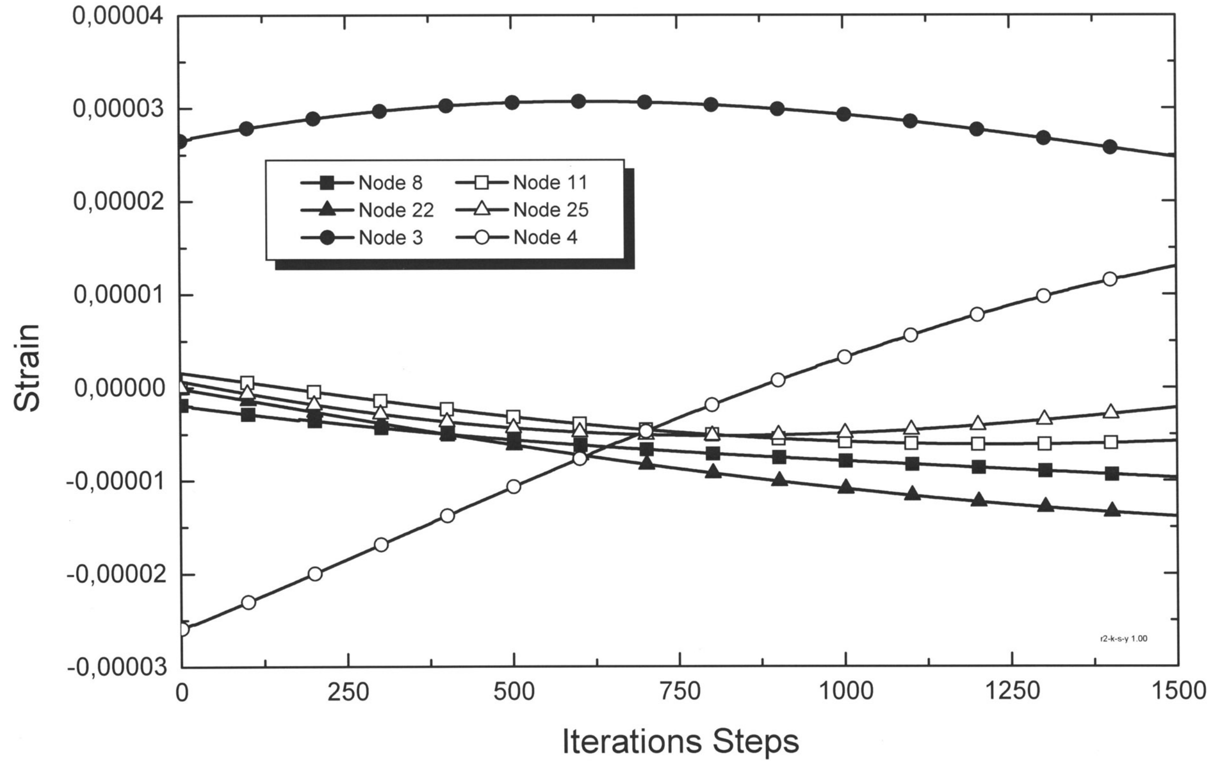

Fig. K4. Course of strain in Y-direction over the iteration steps.

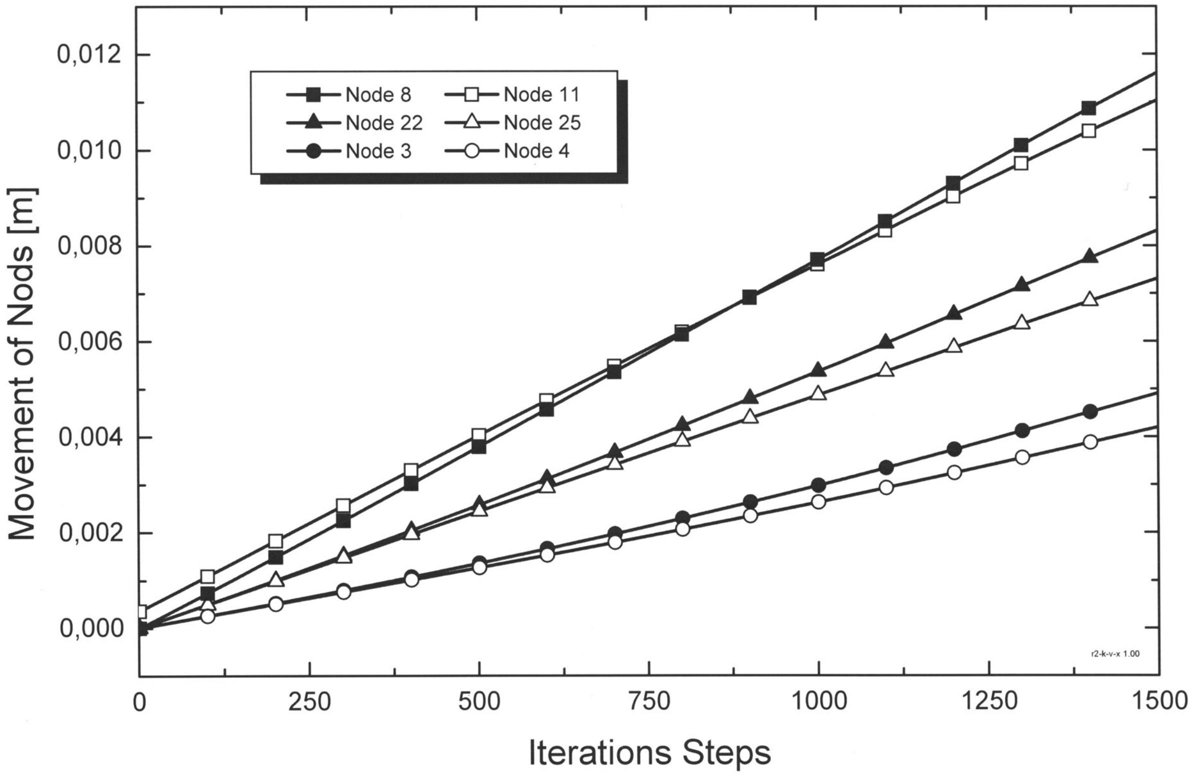

Fig. K5. Course of movement in X-direction over the iteration steps.

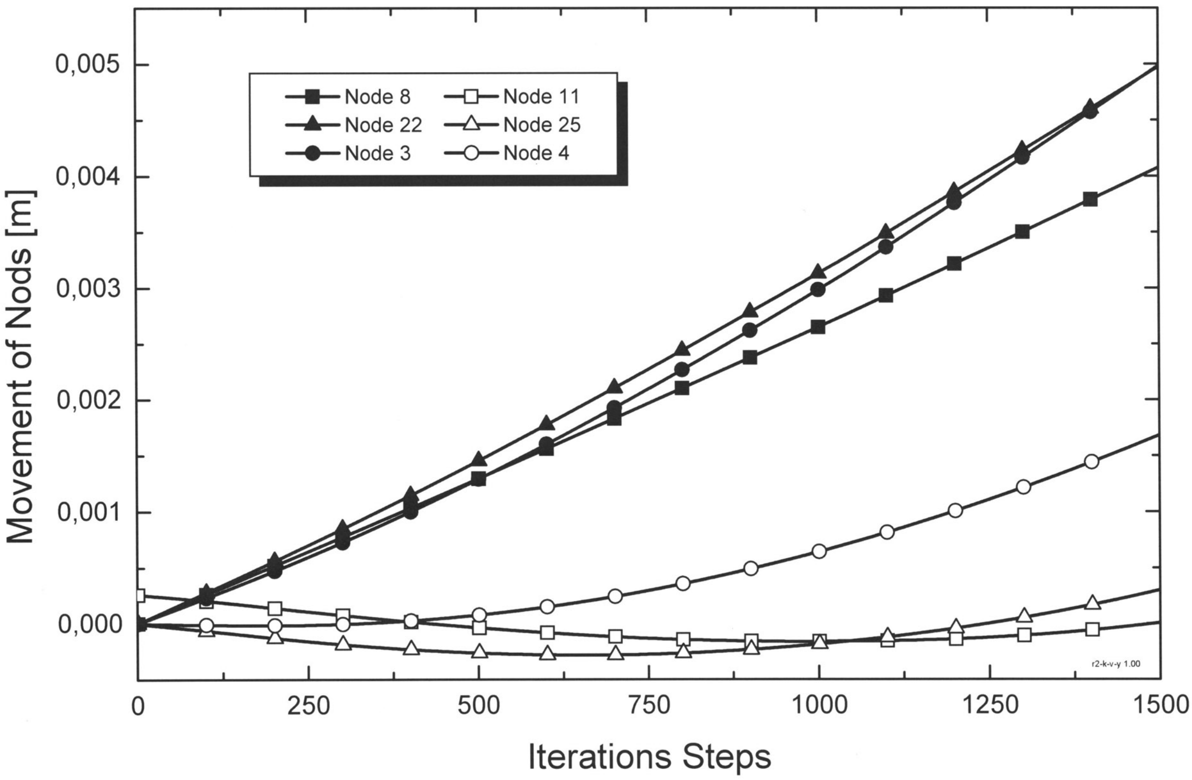

Fig. K6. Course of movement in Y-direction over the iteration steps.

Fig. K7. Radiograph of a molar in comparison with Fig. K2.

Case of tipping: Fx1 = 1 N, Fx2 = 1 N

Here, two equal forces are applied to load the tooth. The force Fx1 is 1 N and the antitipping force Fx2 is also 1 N. The initial centre of rotation (Fig. K1) is situated in the root apex at the beginning of tooth movement. A strong tipping of the tooth during simulation may be expected. After 1500 iterations the centre of rotation has moved from the root to an indefinite position in the alveolar surrounding bone (Fig. K2). This was caused by the considerably changed embedding of the tooth in the jawbone. The local components of strain in X-direction (Fig. K3) indicated that the loadings of most of the nodes in the PDL reduce with increasing iteration number. The jawbone remodels in the way that it can better resist to the force applied. The strain in Y-direction (Fig. K4) indicates that the tooth with larger tipping is more extruded. The movement of the nodes (Fig. K5) shows a linear course, the difference between the individual nodes results from the change of geometry and the strength of the PDL remains almost constant. In Y-direction (Fig. K6) the different movements of the individual nodes arise from the strong tipping movement, too. The entire tipping of the tooth amounts to 24.3 degrees around the Z-axis. Compared to the translation case the Y-component (Fig. K4) of strain is four times larger. The result is a stronger extrusive component. The formation of a pocket and the slight elevation of the jawbone at the left margin may be observed in the case of molars (Fig. K7). Adequate treatment reduces this bone formation to a normal extent again. This model behaviour can be taken as a realistic response of the tooth to the acting force.

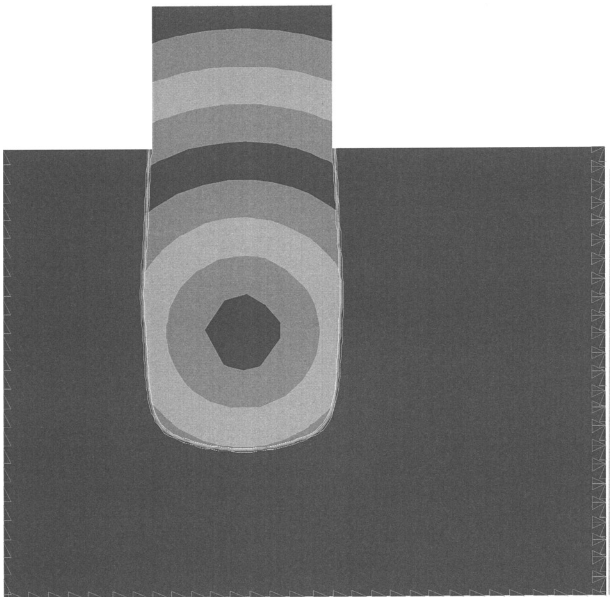

Fig. E1. Starting geometry with initial displacement sum.

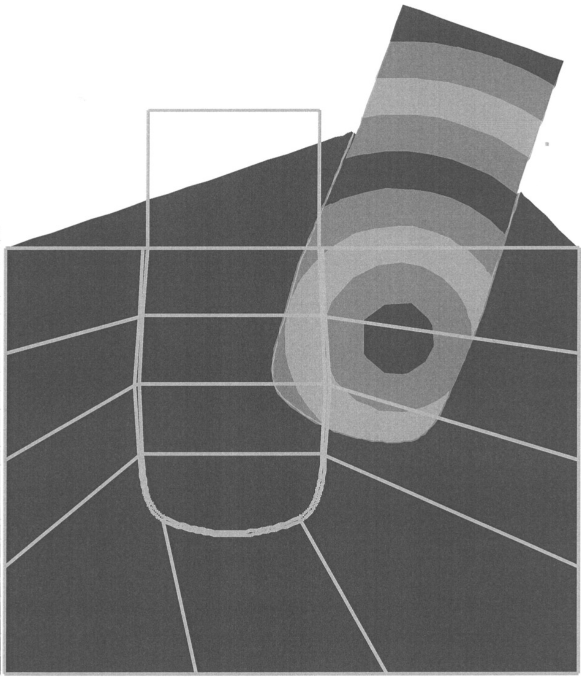

Fig. E2. Geometry and initial displacement sum after 1500 remodelling iterations.

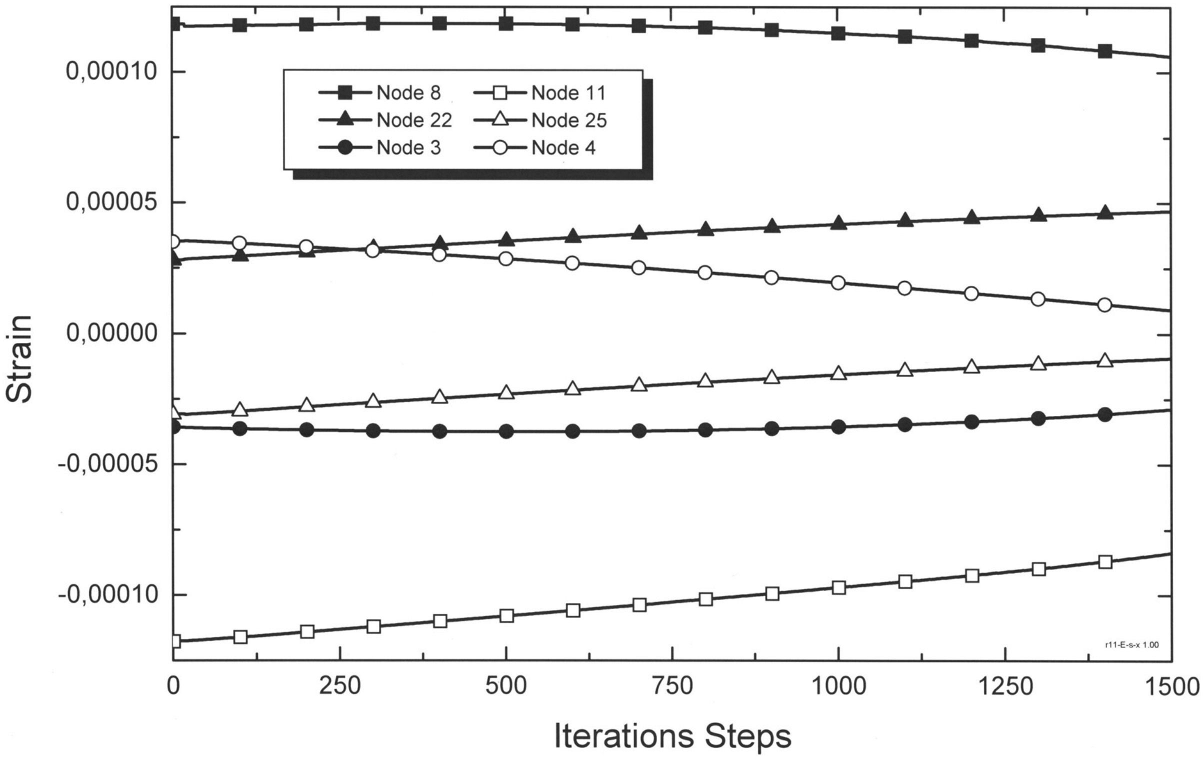

Fig. E3. Course of strain in X-direction over the iteration steps.

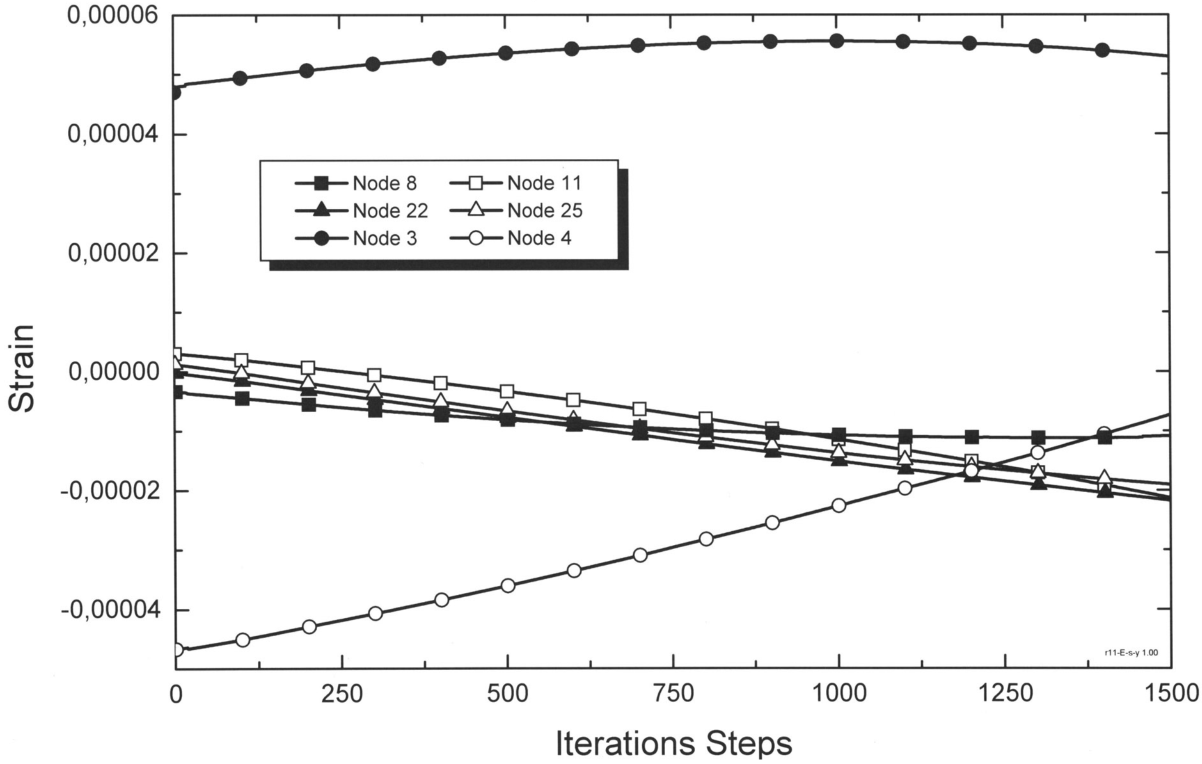

Fig. E4. Course of strain in Y-direction over the iteration steps.

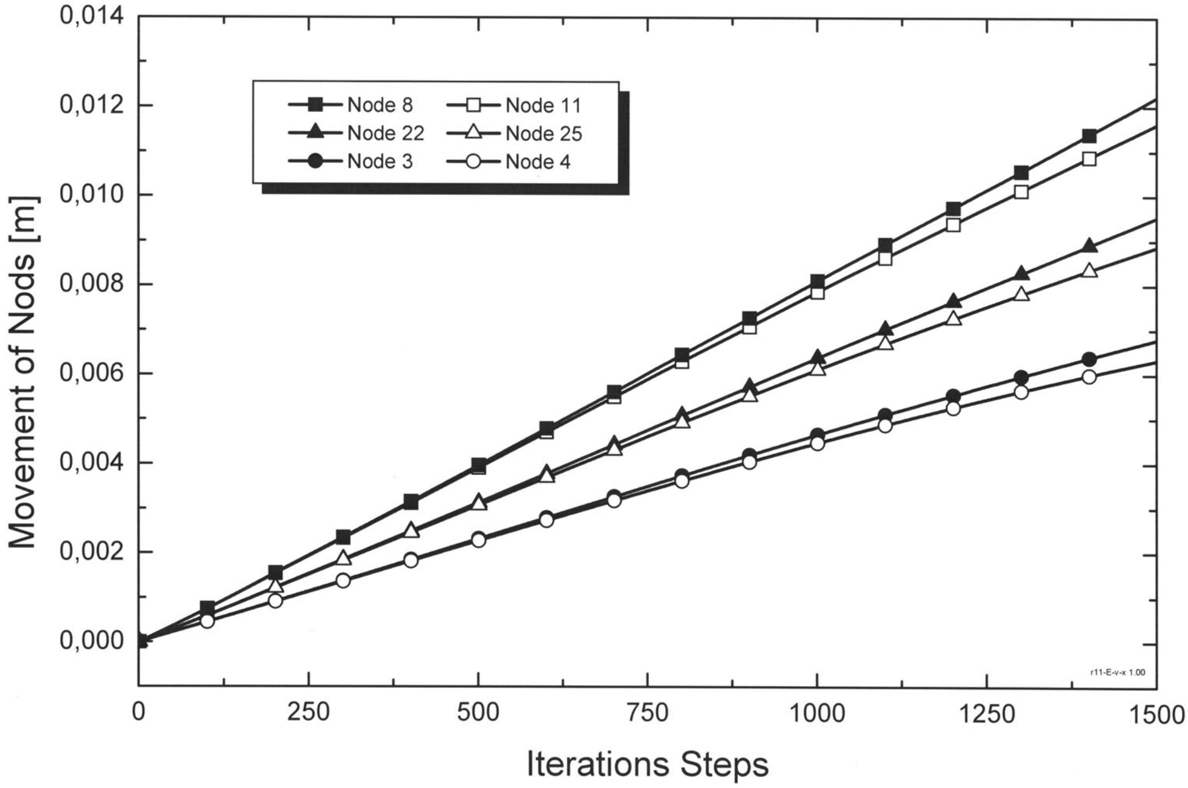

Fig. E5. Course of movement in X-direction over the iteration steps.

Case of extrusion: Fx1 = 1 N

In this case the tooth was loaded with only one force with magnitude Fx1 of 1 N. The centre of rotation is located in the lower third of the tooth root (Fig. E1). Compared to case 2 - the tipping tooth movement - it was to be expected that the centre of rotation is located closer to the force Fx1 in view of the loading. After 1500 iterations it still is on the same level, however, the position is a little bit right of the centre (Fig. E2). This is also an effect of the new geometric embedding of the tooth in the jawbone. Generally, this behaviour of the simulation model is hardly found in practice. The strong growth in Y-direction (Fig. E6) is unnatural for a jawbone. However this effect of decreasing the strain component in X-direction of the PDL and the supradental bone occurs again after several iterations (Fig. E3). The movement in X-direction shows an almost linear course (Fig. E5). The Y-component of strain at node 4 increases with the iteration number (Fig. E4). This increases the bone remodelling in Y-direction.

Summary and Discussion

Generally, these bone remodelling simulations seem to operate well. However the large number of iterations and the fine meshing required in the area of the PDL are problematic. Two cases of translation and tipping presented first will easily be compared qualitatively with clinical cases and a good correspondence can be achieved. Also the behaviour by the iterative calculation, for example the increase of the extrusive component with increasing iteration number, is viewed clinically. In practice it can be observed that teeth easily extrude after a certain period of time in the case of treatment with an inadequate force system. The planned transition to three-dimensional FE-tooth models makes it possible to

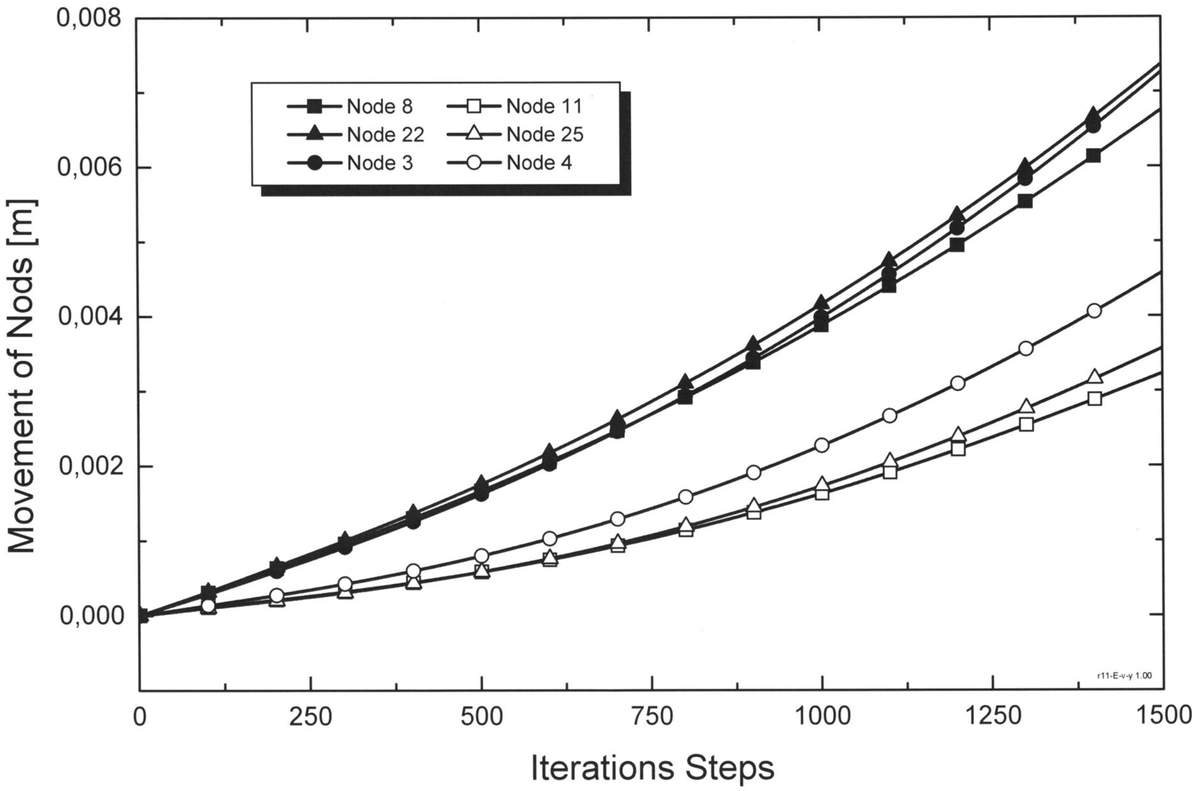

Fig. E6. Course of movement in Y-direction over the iteration steps.

verify the results. In the case of three-dimensional elements, natural boundary conditions can be used for the loads and displacements. The load acting on the tooth is acquired in an suitable measuring set-up by characterising the orthodontic treatment devices concerning their force and moment action [8, 13]. The boundary conditions dependent upon the tooth position then are indicated as optimal load history. The comparison with clinical tooth movements is carried out by means of an optical measuring method which is capable of resolving all movement components, and therefore offers high precision.

The bone remodelling function is optimised and finally tested with the help of a three-dimensional model. The problem with this function is the growth of bone in Y-direction occurring in the case of extrusion. This is unnatural and there is no clinical correspondence. With the remodelling function applied here this problem is conditioned by the model. In principle, it would be possible to construct the model such that the growth in Y-direction on the surface is adjusted after a certain number of iterations. This would represent a permanent bone destruction on the upper side of the tooth. This would mainly influence the results of the calculation, because the tooth would considerably lose its anchorage. A considerably faster tipping and extrusion would be the consequence. It would also be possible to change the bone remodelling algorithm thus that nodes on the surface are treated differently from nodes in the bone. However, this presupposes a remodelling process in the entire volumes of the bony tissue and not only at the boundary layer between alveolus and the PDL. In order to implement this, the biomechanical stimuli should be evaluated in the jawbone and not in the PDL. Furthermore, a more detailed model reconstruction [14] leads to an improvement of the results with the separation of bone material in spongiosa and corticalis with the same remodelling functions. Non-linear material laws [15] and the anisotropy in the PDL probably contribute to another improvement. The periods of calculation increase considerably through the three-dimensional, quite detailed models and the non-linearities.

In summary it is postulated that these two-dimensional calculation models are capable of recording a tooth movement qualitatively. However, for a prediction and an optimisation of treatment, three-dimensional models and patient-specific parameters including the physiological condition must be inserted. The material properties of the PDL may be determined individually by means of *in vivo* measuring methods [15-19]. A non-linear material model has very great effects on bone remodelling. In the model applied, the treatment time correlated with the iteration steps of the simulation. Here, there would be the possibility of correlating the linear bone remodelling parameters with the age in order to achieve a patient-specific adaptation. A detailed statistical comparison with the experiments would show the quality of these functions.

Solutions in the area of the tooth movement with a very coarse but three-dimensional meshing [20] or initial movements were analysed [21]. The system with iterative remodelling function and a remeshing with a reset of geometry still is a novelty. For the first time, it offers the possibility of moving teeth over large distances with a constant mesh quality. This is indispensable because the local analyses require a sufficiently fine meshing. The quality of this function can only be evaluated with three-dimensional models in comparison with the experiments [13]. Then, one would think of a prediction of treatment.

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ПЕРЕСТРОЙКА КОСТИ ПРИ ПЕРЕМЕЩЕНИИ ЗУБА

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Уже в течении длительного времени теории перестройки кости с успехом применяются для описания поведения различных костей скелета человека. Однако эти теории не могут быть непосредственно использованы для анализа ортодонтического перемещения зуба ввиду сложной анатомии пародонта, а также больших изменений геометрии. В настоящей статье решаются частные задачи перестройки кости нижней челюсти с использованием метода конечных элементов. На плоской модели изучаются различные эффекты, имеющие место при перестройке. По возможности проводится сравнение с экспериментальными данными. Вводится функция перестройки, с помощью которой исследуются различные виды перемещения зуба. Библ. 22.

Ключевые слова: первоначальное перемещение зуба, периодонт, метод конечных элементов, свойства материалов, эксперименты *in vivo*

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