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ORIGINAL ARTICLE

Direct versus indirect loading of orthodontic miniscrew implants—an FEM analysis

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Abstract
Objective The mesialization of molars in the lower jaw represents a particularly demanding scenario for the quality of orthodontic anchorage. The use of miniscrew implants has proven particularly effective; whereby, these orthodontic implants are either directly loaded (direct anchorage) or employed indirectly to stabilize a dental anchorage block (indirect anchorage). The objective of this study was to analyze the biomechanical differences between direct and indirect anchorage and their effects on the primary stability of the miniscrew implants.
Materials and methods For this purpose, several computer-aided design/computer-aided manufacturing (CAD-CAM)-models were prepared from the CT data of a 21-year-old patient, and these were combined with virtually constructed models of brackets, arches, and miniscrew implants. Based on this, four finite element method (FEM) models were generated by three-dimensional meshing. Material properties, boundary conditions, and the quality of applied forces (direction and magnitude) were defined. After solving the FEM equations, strain values were recorded at predefined measuring points. The calculations made using the FEM models with direct and indirect anchorage were statistically evaluated.
Results The loading of the compact bone in the proximity of the miniscrew was clearly greater with direct than it was with indirect anchorage. The more anchor teeth were integrated into the anchoring block with indirect anchorage, the smaller was the peri-implant loading of the bone.
Conclusions Indirect miniscrew anchorage is a reliable possibility to reduce the peri-implant loading of the bone and to reduce the risk of losing the miniscrew. The more teeth are integrated into the anchoring block, the higher is this protective effect.
Clinical relevance In clinical situations requiring major orthodontic forces, it is better to choose an indirect anchorage in order to minimize the risk of losing the miniscrew.

Keywords Orthodontics · Miniscrew implants · Anchorage · Molar mesialization · Finite element method (FEM)

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Introduction

The idea of using implants to improve orthodontic anchorage was first published 1945 by Gainsforth and Higley [1]. Kanomi first mentioned titanium miniscrews as temporary anchorage devices in 1997 [2].

Today, miniscrews with diameters of more than 1.2 mm are used universally with success rates of above 70 % [3]. Many factors impact on the success rate [4]. These include the screw design [5–7], screw length [8–12], screw diameter [13, 14], screw angulation [15], screw direction [16], the magnitude of the applied force [17], and the thickness of compact bone [18–20]. A systematic review in 2009 published by Reynders et al. [21], however, revealed a low level of evidence in most studies referring to miniscrew anchorage.

In most surveys, miniscrews were loaded directly [22–24], but due to the relatively high failure rate (15–30 %), some authors suggested the implementation of an indirect anchorage concept [25]. Here, the miniscrew implant is loaded indirectly via an anchorage arch which stabilizes the directly loaded anchor teeth. Until now, no study has been published comparing the biomechanical effects of both miniscrew anchorage concepts. As such, it is not clear which miniscrew concept should be preferred in clinical situations with critical orthodontic anchorage, i.e., mandibular molar mesialization [26].

The finite element method (FEM) is a numerical procedure of applied mathematics. It is a reliable tool for analyzing the biomechanical effects of orthodontic tooth movement [27, 28]. Other authors have used this method to evaluate stress and strain distributions of special palatal orthodontic implants [29], miniscrew implants [30, 31], or orthodontic abutments [32].

As such, the aim of this FEM study was to analyze the biomechanical effects of direct versus indirect loading of miniscrew implants in a typical clinical situation characterized by severe anchorage. Here, two specific questions come to mind. Firstly, will the mode of anchorage influence the stability of the miniscrew and the rate of success? And secondly, should indirect loading of miniscrews be preferred in special clinical situations?

Materials and methods

The nonorganic models (brackets, power arm, anchorage, and segment arches) could be constructed using the software Inventor™ (Autodesk GmbH, Munich, Germany) in a virtual environment using established computer-aided design (CAD) tools. The morphological basis for all the organic models was provided by the coronary layers of a dental CT from a 21-year-old male patient from whom tooth 46 had been extracted 6 years previously. Through manual segmentation of the radiological layers using the software Amira™ (Visage Imaging GmbH, Berlin, Germany), polygon mesh models of the compact and cancellous bone and their corresponding parodontal ligaments could be created for teeth 44, 45, and 47.

CAD modeling

The surface structure of the individual models was then processed afterwards in Rapidform™ (INUS Technology Inc., Seoul, South Korea) and the resulting polygon mesh models were transferred to the Polytrans™ (Okino Inc., Ontario, Canada) program in CAD models. All resulting organic and inorganic CAD models (Fig. 1) were combined in the Mechanical Desktop™ program (Autodesk GmbH, Munich, Germany) with the help of the boolean operations addition and subtraction to form the four simulation models A, B, C, and D (Fig. 1). While simulation models A (without power arm) and B (with power arm) represented the direct application of force on the miniscrew (direct anchorage), with simulation models C and D, the gap closure was carried out by indirect anchorage of the anchor teeth using the miniscrew (Fig. 1).

FEM modeling

The resulting simulation models could now be imported as geometry into the ANSYS™ 11.0 program (Ansys Inc., Canonsburg, PA, USA) before being meshed three-dimensionally into a finite element model in which the individual elements consisted of parabolic tetrahedrons. Thresholds and limits were defined according to appropriate material properties as taken from the literature (Table 1). Into each simulation model, a reciprocal force vector (1.5 N) was introduced, where the three-dimensional orientation of the force vectors is illustrated in Fig. 1. The boundary conditions and thresholds for all four simulation models were identical and were defined over several fixed nodes at the edge of the model. The contact conditions between all structures were defined as a group, with exception of the contact between the bone and the miniscrew, which was defined as friction-affected (coefficient μ=0.3). After completion of the simulation calculations, the induced relative strain values (in microstrain; μstrain) could be recorded with the help of an interactive measuring tool at defined measurement points of the peri-implant bone.

Statistical analysis

Using the software SPSS™ 15.0 (SPSS Inc., Chicago, IL, USA) the measured values were statistically evaluated (Tables 2 and 3). For detecting significant differences between individual groups of measured values, the Wilcoxon test for dependent samples and the Mann–Whitney U test for independent samples were used.

Results

Upon direct traction of the miniscrew toward bracket 47 (simulation A), the highest strain values were observed in the upper third of the peri-implant bone. At the measurement points within the area of the cortical bone, peak values greater than 60 μstrain could be measured. During simulation A, the average value for all measuring points (n=12) was 26 μstrain (SD, 24 μstrain) and lay in the area of the compact bone (Table 2 and Fig. 4). For the measuring points in the area of the cancellous (Table 3 and Fig. 5) bone (n=24), these values lay at 5 μstrain (SD, 2 μstrain). The Mann–Whitney U test for independent samples showed a highly significant difference (p<0.001) between the measured values for simulation A in the areas of the compact and the cancellous bone.

If the force vector ran from the miniscrew in the direction of the power arm of bracket 47 (Figs. 2 and 3), then the highest strain values could also be measured in the upper third of the peri-implant bone. The peak values ranged here

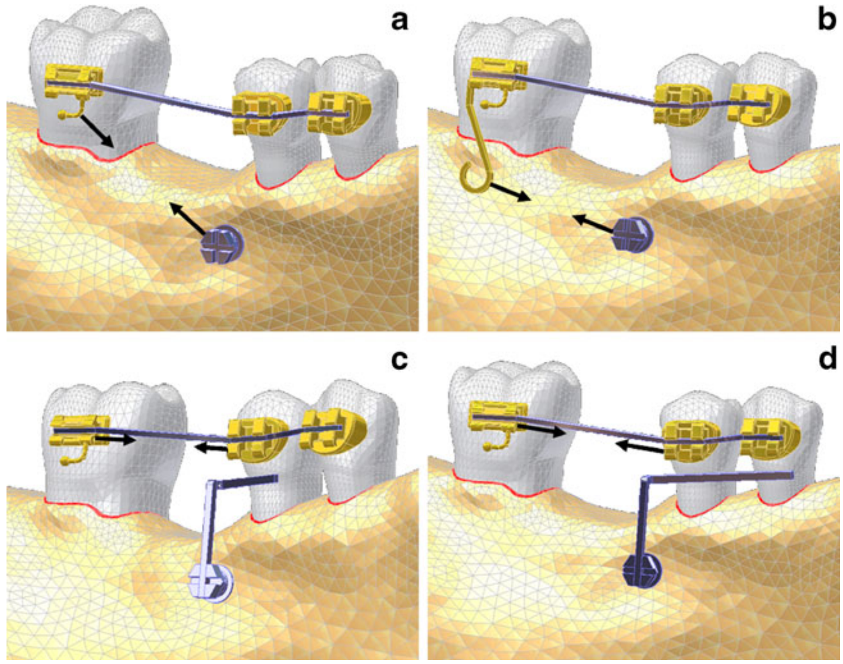

Fig. 1 Composed CAD models used in simulations A (**a**), B (**b**), C (**c**), and D (**d**). The direction of applied forces is illustrated by vectors

from 60 µstrain (caudal to the miniscrew) to 83 µstrain (distal to the miniscrew). The average value for all measuring points (Table 2) in the area of the compact bone (Fig. 4; n=12) was 30 µstrain (SD, 30 µstrain). For the measuring points in the area of the cancellous bone (n=24), these values (Table 3 and Fig. 5) lay at 5 µstrain (SD, 2 µstrain). For simulation B, the Mann–Whitney U test for independent samples also showed a highly significant difference ($p<$ 0.001) between the measured values in the areas of the compact and the cancellous bone.

In simulation C (indirect anchorage of tooth 45), the situation was different. Here, a more homogeneous distribution of measured values was observed for all peri-implant measuring points. The maximum value was 12 µstrain which was measured in the apical third mesial to the miniscrew. The average value for all measuring points (Table 2) in the area of the compact bone (Fig. 4; n=12) was 4 µstrain (SD, 3 µstrain).

For the measuring points in the area of the cancellous bone (Table 3 and Fig. 5; n=24), these values lay low as 5 µstrain (SD, 2 µstrain). The difference between the measured values in the compact and cancellous bone proved to be significant by Mann–Whitney U test (p=0.013).

In the same way as could be seen with simulation C, a similar situation was presented with simulation D regarding the distribution of the measured values, where the miniscrew served as an indirect anchorage for teeth 45 and 44 (Figs. 2 and 3). The maximum value was 8 µstrain which was measured in the apical third mesial to the miniscrew in the peri-implant bone. The average value for all measuring points (Table 2; n=12) in the area of the compact bone (Fig. 4) was 3 µstrain (SD, 2 µstrain). For the measuring points in the area of the cancellous bone (Table 3 and Fig. 5; n=24), these values lay at 4 µstrain (SD, 2 µstrain). Simulation D as well the Mann–Whitney U test revealed a significant difference (p=0.015) between the measured values in the compact and cancellous bone.

Significant differences (Wilcoxon test for dependent samples) between the individual simulations could be found

Table 1 Material properties used in the simulations A, B, C, and D: Young's modulus is specifying the elasticity and Poisson's ratio is specifying the transverse contraction characteristics of the material

Morphology	Young's modulus (GPa)	Poisson's ratio
Teeth	22.00	0.31
PDL	0.069	0.45
Brackets	193.0	0.35
Segmental arch	193.0	0.35
Anchorage arch	193.0	0.35
Miniscrew	110.0	0.35
Compact bone	13.70	0.33
Cancellous bone	1.370	0.30

Table 2 Descriptive statistics of measured strain values (microstrain) in compact bone (n=12)

Compact bone	A	B	C	D
n	12	12	12	12
Mean	26	30	4	3
Median	13	13	3	2
SD	24	30	3	2
Minimum	4	5	2	1
Maximum	65	83	11	6
Range	61	78	9	5

Table 3 Descriptive statistics of measured strain values (microstrain) in cancellous bone (n=24)

Cancellous bone	A	B	C	D
n	24	24	24	24
Mean	5	5	5	4
Median	5	5	5	3
SD	2	2	2	2
Minimum	2	2	2	2
Maximum	9	12	12	8
Range	7	10	10	6

in the area of the compact bone. Here, highly significant differences (p=0.002) were shown between simulations A and C, A and D, B and C, B and D, as well as C and D. However, the differences between simulations A and B did not prove to be significant. Also, in the area of cancellous bone, no significant differences could be found.

Discussion

Systematic error

The FEM is a suitable instrument for investigating biomechanical problems [27, 32–34]. The systematic error of these numeric computations decreases along with the precision with which the models used can reproduce the morphology and material properties of the real anatomical structure [35]. For the specific analysis of the biomechanics of miniscrews, FEM models have been presented which only roughly reproduce reality regarding anatomic morphology and the clinical apparatus used [16, 29, 31]. As such, the bone segments were not modeled from a patient, but were instead arbitrarily constructed. This schematization can result in a relatively high systematic error for the calculations [32, 35]. In order to reduce this systematic error, both the patient's individual anatomy and the clinical treatment situation (including the use of orthodontic arches and brackets) were taken into account in this study. One disadvantage of the FEM models used in this study was the fact that the material properties of the bony structures, just as was the case in the studies of Chen, Pickard, and Lombardo, were defined in simplified form as linear and isotropic [16, 29, 31]. A detailed representation of the complex bone material properties (anisotropy and nonlinearity) was refrained to economize on time and calculations as optimally as possible. However, the systematic error arising from this simplification was reduced by the fact that it was not the absolute measurements that were used for providing information, but much rather the differences between the individual simulations. In this way, the informative power of the findings was in fact increased. As such, the objective of this study was not to determine precise strain values for a particular clinical situation, but much rather to evaluate the extent and significance of deviations between different clinical situations.

Direct anchorage

From the results of this study, we can state that the peri-implant bone within the area of the compact bone is substantially more loaded with direct than it is with indirect anchorage. Since the area of the compact bone is important for the primary stability of the miniscrews [36], the results

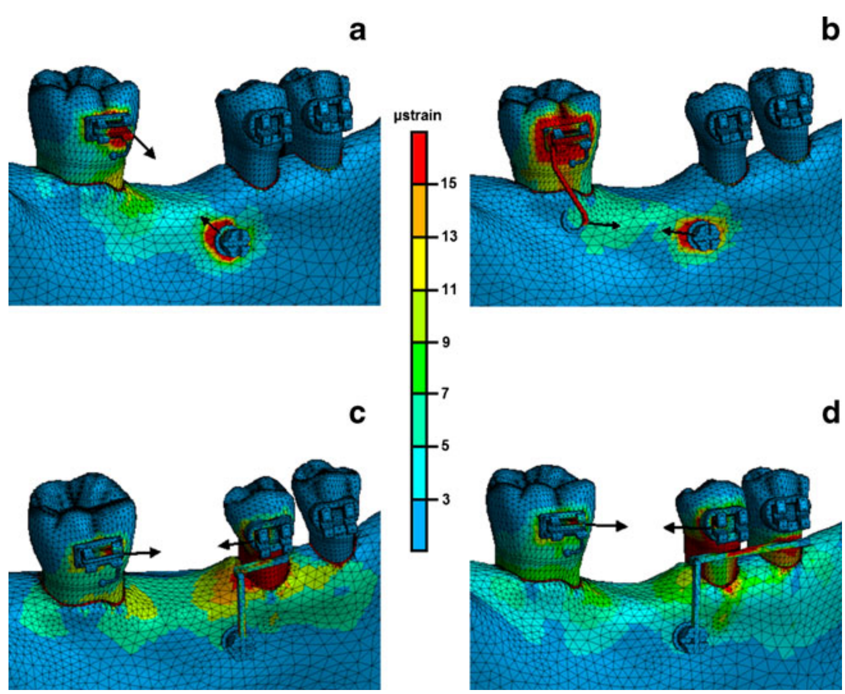

Fig. 2 Comparison of induced strain values (in microstrain) in simulations A (**a**), B (**b**), C (**c**), and D (**d**)

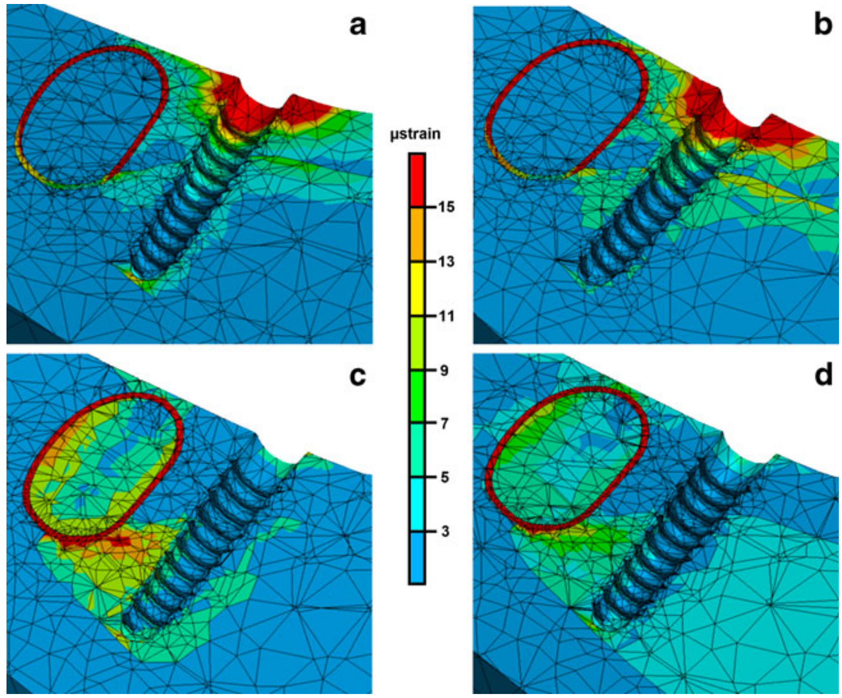

Fig. 3 Comparison of induced strain values (in microstrain) in simulations A (**a**), B (**b**), C (**c**), and D (**d**) (horizontal cross section in the bony region near the miniscrew, which was faded out)

obtained might provide a good explanation for the relatively high loss rates with direct anchorage of the miniscrews [3]. Higher forces are needed with mesialization of lower jaw molars in particular, which with direct force action on the miniscrew can lead to high loads on the peri-implant compact bone, which in turn could result in displacement [37] or a loosening of the miniscrew [38]. According to our results, it is irrelevant whether the force is applied directly to the bracket of the tooth to be moved or at a powerarm. Various authors have already reported on how important an adequate anchorage of the miniscrew in the compact zone of the bone is for primary stability, since no osseointegration of the screw can be expected [39, 40]. Also, the superiority of bicortical screw anchorage compared to a monocortical anchorage [13, 41] emphasizes the role of the compact bone for a successful orthodontical treatment using miniscrews [18–20]. The cancellous bone on the other hand is less important for primary stability and the prognosis of the miniscrews [39, 42, 43]. As such, the results also show no significant differences between direct and indirect anchorage of the miniscrews for cancellous bone. In the case of indirect anchorage, the measured strain values in the peri-implant bone were much lower than they were with direct anchorage. This offers a good explanation for the clinical finding that the risk of losing miniscrews is much smaller with this type of anchorage [25, 44].

Indirect anchorage

In the case of indirect anchorage, however, significant differences were apparent for the number of anchor teeth

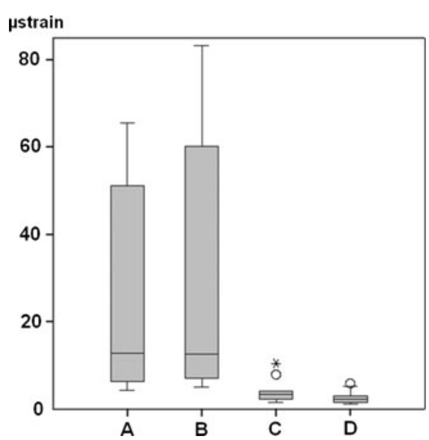

Fig. 4 Statistical relation of strain values (compact bone) in simulations *A*, *B*, *C*, and *D*

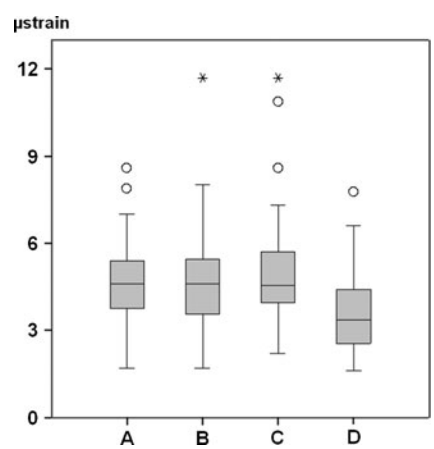

Fig. 5 Statistical relation of strain values (cancellous bone) in simulations *A*, *B*, *C*, and *D*

involved. The more teeth that were integrated into the anchorage block, the smaller was the loading of the peri-implant compact bone, and in turn the smaller was the risk of miniscrew loss, which was in close agreement with our own clinical observations. With a complicated anchorage scenario, as presented by mesialization of a lower jaw molar, it is advisable to select an indirect anchorage and to integrate as many teeth as possible into this anchorage block to keep the risk of losing the miniscrew to a minimum. The blocking between anchor teeth and miniscrew should be very rigid in order to protect the anchor teeth from any unwanted movements.

Conclusions

Obviously, the loading of the compact bone in the proximity of the miniscrew is greater with direct than with indirect anchorage. The more anchor teeth that are integrated into the anchoring block with indirect anchorage, the smaller is the peri-implant loading of the bone. In clinical situations requiring major orthodontic forces, it is better to choose an indirect anchorage in order to minimize the risk of losing the miniscrew. In this context, the anchoring teeth should be fixed to the miniscrew rigidly.

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