Finite element analysis of mono- and bicortical mini-implant stability

Christof Holberg, Philipp Winterhalder, Ingrid Rudzki-Janson and Andrea Wichelhaus
Department of Orthodontics, University of Munich, Bavaria, Germany

Correspondence to: Christof Holberg, Department of Orthodontics, University of Munich, Goethestrasse 70, 80336 Munich, Germany. E-mail: cholberg@med.lmu.de

SUMMARY
OBJECTIVES Loosening and loss rates of monocortical mini-implants are relatively high, therefore the following null hypothesis was tested: ‘The local bone stress in mono and bicortically-anchored mini-implants is identical’.

MATERIAL AND METHODS Anisotropic Finite Element Method (FEM) models of the mandibular bone, including teeth, periodontal ligaments, orthodontic braces, and mini-implants of varying length, were created. The morphology was based on the Computed Tomography data of an anatomical preparation. All mini-implants with varying insertion depths (monocortical short, monocortical long, bicortical) were typically loaded, and the induced effective stress was calculated in the cervical area of the cortical bone. The obtained values were subsequently analysed descriptively and exploratively using the SPSS 19.0 software.

RESULTS The null hypothesis was rejected, since the stress values of each anchorage type differed significantly (Kruskal–Wallis Test, \( P < 0.001 \)). Therefore, the lowest effective stress values were induced in bicortical anchorage (mean = 0.65 MPa, SD = 0.06 MPa) and the highest were induced in monocortical (short) anchorage of the mini-implants (mean = 1.79 MPa, SD = 0.29 MPa). The Spearman rank correlation was \( 0.821 (P < 0.001) \).

CONCLUSIONS The deeper the mini-implant was anchored, the lower were the effective stress values in the cervical region of the cortical bone. Bicortical implant anchorage is biomechanically more favourable than monocortical anchorage; therefore, bicortical anchorage should be especially considered in challenging clinical situations requiring heavy anchorage.

Introduction

To avoid undesired effects on the adjacent teeth, sufficient anchorage should be ensured during orthodontic tooth movement (Angle, 1899). In addition to dental anchorage, where multiple teeth can be assembled to anchor blocks, skeletal anchorage is a proven method, particularly in challenging clinical situations (Feldmann and Bondemark, 2006). For example, the anchorage system is strongly loaded during mandibular molar mesialization, as in such cases either an extraoral (Benauwt, 1974) or a skeletal anchorage system is strongly recommended (Freudenthaler et al., 2001). For skeletal anchorage, dental implants (Gallas et al., 2005), mini-plates (Sato et al., 2007; Cornelis et al., 2008), palatal implants (Gedrange et al., 2005; Wehrbein and Gollner, 2007), and mini-implants (Kanomi, 1997) are used. Mini-implants have been particularly favoured for skeletal anchorage in recent years (Reynders et al., 2009). However, the loss and loosening rates of these usually monocortically inserted mini-implants is relatively high at 10–30% (Crismani et al., 2010). Many factors impact on the success rate (Stahl et al., 2009; Chang et al., 2012). These include the direction (Pickard et al., 2010) and the magnitude of the applied force, the thread design (Wiechmann et al., 2007), the thread diameter (Morarend et al., 2009), and above all, the length of the mini-implant (Mortensen et al., 2009). The chosen screw length of the mini-implant determines the kind of anchorage (mono- or bicortical). Thus monocortical anchorage obviously reduces the risk of injury to the adjacent anatomic structures such as periodontal ligaments (PDLs) (Lemieux et al., 2011). Longer mini-implants, however, could probably enhance primary stability, leading to a reduced risk of loosening or loss (Wu et al., 2007), even if the injury risk to adjacent anatomical structures may be increased (Lemieux et al., 2011). Thus the factors of required anchorage and injury risk should be considered when choosing a specific mini-implant length (Sung et al., 2010; Çifter and Saraç, 2011). In challenging clinical situations requiring heavy anchorage, the insertion of bicortically anchored mini-implants comes into consideration. Since a detailed biomechanical comparison between mono- and bicortical anchorage systems is not available in literature as yet; this will be the subject of the present study. In particular, the following questions will be answered. Are there biomechanical differences between a
MONO- OR BICORTICAL MINI-IMPLANTS

monocortical short, a monocortical long, and a bicortical insertion of the mini-implants? Are the biomechanical differences of importance? Do the differences affect the quality of the primary stability? Thus, the null hypothesis was, ‘The load of the peri-implant bone in mono or bicortical anchorage is identical’.

Material and methods

An anatomical preparation of a mandibular segment (18 year old man) with tooth 46 missing served as the morphological basis for the Finite Element Method (FEM) models (Figure 1). This anatomical specimen was scanned using Computed Tomography (CT), and the morphological data were acquired by manual segmentation of the axial CT layers using the Amira™ software (Visage Imaging GmbH, Berlin, Germany). Segment models of the compact and cancellous mandible, complete models of tooth 44, 45, and 47, and models of the corresponding PDLs resulted after cross-linking the point cloud (Delauney Triangulation) to three-dimensional polygon meshes. The surface structure of the polygon meshes was subsequently post-processed and optimized using the Rapidform™ software (INUS Technology Inc., Seoul, Korea).

Pre-processing

Using Polytrans™ software (Okino Inc., Ontario, Canada), the polygon meshes were converted into analytical solid models (IGES format). All constructible elements (brackets, powerarm and mini-implant) were designed virtually using established Computer Aided Design (CAD) tools in the Inventor™ software (Autodesk GmbH, Munich, Deutschland). The resulting solid models could be combined to three compound models (Figure 2) using Boolean operations of addition and subtraction in the Mechanical Desktop™ software (Autodesk GmbH, Munich, Germany). The first CAD model represented short monocortical anchorage by using a mini-implant with a diameter of 1.6 mm and a length of 5 mm. The mini-implant in the second model (7 mm length) was also inserted monocortically but reached the region of the cancellous bone. Finally, in the third CAD model bicortical anchorage was performed by using a mini-implant with a diameter of 1.6 mm and a length of 10 mm. The resulting CAD models could now be cross-linked three-dimensionally to FEM models using the ANSYS™ 11.0 simulation software (Ansys Inc., Canonsburg, Pennsylvania, USA). The finite elements used were parabolic tetrahedrons with one node at each corner and one additional node in the middle of each edge. Accordingly, this tetrahedral element was a nonlinear, 10-noded three-dimensional finite element. Loading and boundary conditions (Tables 1 and 2) were defined corresponding to the standard values found in the literature. A force and a counterforce each of 1.5 N was applied to all simulation models. The orientation of the force vector corresponded to the closed coil depicted in Figure 1. As a boundary condition, multiple nodes could be fixed in the simulation model as far away as possible from the region of interest (ROI). The contact conditions

Figure 1 CAD model consisting of various materials (cortical bone, cancellous bone, teeth with enamel, dentin and periodontal ligament, brackets, powerarm, and a mini-implant with a variable length).

Figure 2 Cross-section of the CAD models showing the variable length of the mini-implant that was anchored mono- or bicortically.
between the structures were defined as fixed—only the contact between the bone and the mini-implant was friction dependent with non-linear lifting (coefficient $\mu = 0.3$).

**Stress calculation**

An ROI was defined, where the effective stress (in MPa) should be calculated. This ROI was formed by the buccal part of the cortical bone surrounding the head of the mini-implant by a diameter of 8mm (Figure 3). In this region the effective stress (in MPa) was calculated at all nodes. Following this, the values were sorted by their size, and the top 1000 highest stress values were considered for further statistical analysis, as these peak loads could be responsible

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**Table 1**  Material properties of the mini-implant and other materials used in the present study; Young’s modulus specifies the elasticity and Poisson’s ratio the transverse contraction characteristics of the material.

<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mini-implant</td>
<td>110 000</td>
<td>0.35</td>
<td>(Motoyoshi et al., 2009)</td>
</tr>
<tr>
<td>Brackets</td>
<td>210.0</td>
<td>0.30</td>
<td>(Chen et al., 2008)</td>
</tr>
<tr>
<td>Adhesive</td>
<td>8.823</td>
<td>0.25</td>
<td>(Lin et al., 2011)</td>
</tr>
<tr>
<td>Powerarm (Stainless Steel)</td>
<td>193.0</td>
<td>0.30</td>
<td>(Chen et al., 2008)</td>
</tr>
<tr>
<td>Enamel</td>
<td>80 000</td>
<td>0.25</td>
<td>(He et al., 2006)</td>
</tr>
<tr>
<td>Dentin</td>
<td>24 400</td>
<td>0.43</td>
<td>(Xu et al., 1998; Kinney et al., 2004)</td>
</tr>
<tr>
<td>PDL</td>
<td>50</td>
<td>0.49</td>
<td>(Rees and Jacobsen, 1997; Fill et al., 2011)</td>
</tr>
<tr>
<td>Nerve tissue</td>
<td>0.58</td>
<td>0.42</td>
<td>(Ichihara et al., 2001; Borschel et al., 2003)</td>
</tr>
</tbody>
</table>

**Table 2**  Anisotropic material properties of cortical and cancellous bone used in the present study (Dechow et al., 1993; O’Mahony et al., 2000; O’Mahony et al., 2001).

<table>
<thead>
<tr>
<th></th>
<th>Compact Bone</th>
<th>Cancellous Bone</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_x$</td>
<td>12 600 MPa</td>
<td>1148 MPa</td>
</tr>
<tr>
<td>$E_y$</td>
<td>19 400 MPa</td>
<td>1148 MPa</td>
</tr>
<tr>
<td>$E_z$</td>
<td>12 600 MPa</td>
<td>210 MPa</td>
</tr>
<tr>
<td>$G_{xy}$</td>
<td>5700 MPa</td>
<td>434 MPa</td>
</tr>
<tr>
<td>$G_{yz}$</td>
<td>5700 MPa</td>
<td>68 MPa</td>
</tr>
<tr>
<td>$G_{xz}$</td>
<td>4850 MPa</td>
<td>68 MPa</td>
</tr>
<tr>
<td>$v_{xy}$</td>
<td>0.253</td>
<td>0.322</td>
</tr>
<tr>
<td>$v_{yz}$</td>
<td>0.390</td>
<td>0.055</td>
</tr>
<tr>
<td>$v_{xz}$</td>
<td>0.300</td>
<td>0.055</td>
</tr>
</tbody>
</table>

**Figure 3**  Comparison of induced stress values (in MPa) in the region of interest (cortical bone around the head of the mini-implant). The implant body itself was removed for a better view. Cross-section (left row) showing the distribution of the induced effective stresses (in MPa).
for possible mini-implant complications such as loss of the primary stability, loosening, or even loss of the mini-implant.

**Statistical analysis**

The obtained values were analysed descriptively and exploratively using the SPSS™ 15.0 software. The Kolmogorov–Smirnov test was performed to check the variables for normal distribution. The Kruskal–Wallis test was used to find significant differences between the samples; following this, the Mann–Whitney U-test was used in pairs. The correlation coefficient according to Spearman was calculated to quantify the correlation between the length of the mini-implant and the induced effective stress in the peri-implant bone.

**Results**

The samples for the anchorage variants ‘monocortical short’, ‘monocortical long’, and ‘bicortical’ differed with regard to their descriptive statistics. The mean effective stress (in MPa) was clearly higher in mono than in bicortical anchorage (Figure 3). The peri-implant stress values in the cortical bone for the monocortical variant were more than twice those for the bicortical variant. The individual values, which were sorted in a descending order, are depicted in Figure 4 to give a comparison between the anchorage types.

**Monocortical anchorage**

After applying a force of 1.5 N perpendicular to the axis of the monocortical short-inserted mini-implant, the mean effective stress in the cortical bone was $1.79 \text{ MPa}$ (SD $0.29 \text{ MPa}$). The median was $1.71 \text{ MPa}$, the maximum value $2.76 \text{ MPa}$, and the minimum value $1.43 \text{ MPa}$. The long variant of the monocortical mini-implant showed somewhat lower stress values. The mean effective stress was $1.55 \text{ MPa}$ (SD $0.28 \text{ MPa}$). The median was $1.48 \text{ MPa}$, the maximum $2.43 \text{ MPa}$, and the minimum $1.22 \text{ MPa}$. (Table 3).

**Bicortical anchorage**

After applying a force of 1.5 N perpendicular to the axis of the monocortical short-inserted mini-implant, the mean effective stress in the cortical bone was $0.65 \text{ MPa}$ (SD 0.09 MPa). The median was 0.64 MPa, the maximum value 0.90 MPa, and the minimum 0.57 MPa (Table 3). The box plots in Figure 5 illustrate the statistical differences between the three mini-implant anchorage variants.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Unit</th>
<th>Monocortical (short)</th>
<th>Monocortical (long)</th>
<th>Bicortical</th>
</tr>
</thead>
<tbody>
<tr>
<td>$n$</td>
<td></td>
<td>1000</td>
<td>1000</td>
<td>1000</td>
</tr>
<tr>
<td>Mean</td>
<td>MPa</td>
<td>1.79</td>
<td>1.55</td>
<td>0.65</td>
</tr>
<tr>
<td>Median</td>
<td>MPa</td>
<td>1.71</td>
<td>1.48</td>
<td>0.64</td>
</tr>
<tr>
<td>Std.-Dev.</td>
<td>MPa</td>
<td>0.29</td>
<td>0.28</td>
<td>0.06</td>
</tr>
<tr>
<td>Minimum</td>
<td>MPa</td>
<td>1.43</td>
<td>1.22</td>
<td>0.57</td>
</tr>
<tr>
<td>Maximum</td>
<td>MPa</td>
<td>2.76</td>
<td>2.43</td>
<td>0.90</td>
</tr>
</tbody>
</table>

**Table 3** Descriptive statistics of effective stress values (n=1000) calculated in the cervical region of the cortical bone close to the mini-implant.

**Figure 4** Comparison of the peak stress values with different types of anchorage calculated at the nodes of the region of interest (cortical bone around the head of the mini-implant). In the chart all presented nodes were rearranged by their value size.

**Figure 5** Boxplot visualization comparing the stress values in the region of interest (cortical bone around the head of the mini-implant) with three different types of mini-implant anchorage.
with a very high significance ($P < 0.001$). Due to the lack of normal distribution, non-parametric tests were used. In the Kruskal–Wallis test the mean values of the samples differed with high significance ($P < 0.001$). A pairwise testing of the samples using the Mann–Whitney U-test yielded highly significant differences for each variant ($P < 0.001$). The rank correlation according to Spearman was in the negative range at $-0.821$ ($P < 0.001$), which corresponds to a high correlation. When only including the 10 highest stress values in the samples, the correlation coefficient according to Spearman was $-0.943$ ($P < 0.001$), which corresponds to a very high correlation. The negative correlation coefficients indicated that the peri-implant stress values were lower when the mini-implant was longer and inserted deeper into the alveolar bone.

**Discussion**

The FEM is a proven method to analyse mini-implant-based anchorage systems biomechanically (Motoyoshi et al., 2005; Stahl et al., 2009; Chatzigianni et al., 2011; Ammar et al., 2011; Suzuki et al., 2011; Woodall et al., 2011; Liu et al., 2012; Singh et al., 2012). In the present study, three anchorage variants of mini-implants were compared biomechanically using the FEM. As in other FEM studies, the FEM models represented an idealization and simplification of the reality (Cattaneo et al., 2005; Stahl et al., 2009; Pickard et al. 2010; Chatzigianni et al., 2011). For example, the mechanical properties of the PDL are very complex in reality (Poppe et al., 2002). Different approaches are taken to consider this complexity in FEM studies; for instance, sometimes the material properties of the PDL are defined as linear (Rees and Jacobsen, 1997; Poiate et al., 2009; Fill et al., 2011), but this approach often seems to be an oversimplification. Often the material properties are defined as ‘bilinear’, leading to an ‘ultimate strain’ value between the linear ranges (Ziegler et al., 2005; Cattaneo et al., 2009; Dong-Xu et al., 2011). This approach is more accurate because both ‘curled’ and ‘uncurled’ PDL modes are taken into account (Ziegler et al., 2005; Dong-Xu et al., 2011). Another approach is the description of the PDL by the hyper-elastic Ogden constitutive law (Shibata et al., 2006). In the present study, the material properties of the PDL were defined as linear. This simplification was acceptable, as the PDL was far away from the ROI and had only a small effect on the results. In the present study, the morphology of the bone was differentiated into a cortical and cancellous segment, and the corresponding anisotropic material properties could be assigned. All the material properties used were taken from the literature. In comparison to other FEM studies, the morphological accuracy of the present FEM model was relatively high, because the alveolar bone segments were sometimes virtually constructed by the authors and not transferred from individual patient data (Pickard et al., 2010; Lombardo et al., 2010). This virtual design of organic structures increases the methodical error. To keep this as low as possible, the FEM models of the present study were not constructed but rather generated from the CT data of an anatomical preparation. To further reduce the systematic error, no absolute values were considered to draw the conclusion, only the differences between the simulations. Since all simulations were affected by the simplification effects to the same extent, the analysis of the differences resulted in an additional increase of validity.

**Cortical stress**

The null hypothesis (‘The local bone stress in mono and bicortically-anchored mini-implants is identical’) could be discarded since highly significant differences were determined between the different anchorage types. Since in bicortical anchorage the peak stresses induced in the cortical bone were the lowest, this type of anchorage seems to be superior according to a long-lasting primary stability. Respectively, the risk of mini-implant loss or loosening should be reduced in bicortical anchorage. The present results are supported by the study of Brettin et al. (2008) in which the deflection forces of mono or bicortically anchored mini-implants were calculated using a simplified FEM model and an anatomical preparation. Here the bicortically anchored mini-implants underwent less loosening than the monocortical ones. Additionally, biomechanical differences between variants of monocortical anchorage could be analysed in the present study. Since the short monocortical anchorage is limited to the outer cortex, the stresses were higher than with longer mini-implants, which are extended into the cancellous bone. The stress differences between these monocortical variants were highly significant, but the extent of the differences was much lower than between monocortical and bicortical anchorage.

**Monocortical anchorage**

For a sustainable primary stability of orthodontic mini-implants it is important to prevent loosening of the mini-implant by overloading the peri-implant bone. In particular, the stress induced in the cervical region of the peri-implant bone is important here (Singh et al., 2012). The lower the induced stress is here, the lower the risk of loosening of the mini-implant (Florvaag et al., 2010). With short monocortical anchorage, however, relatively high stresses are induced in the region of the cervical peri-implant bone. These were somewhat higher than in long monocortical anchorage but clearly higher than in bicortical anchorage. Not only was the peri-implant bone in the distal region of the mini-implant affected by stresses (where the tensile force was applied by the closed coil) so too was the mesial region. Obviously, the head of the short mini-implant was pulled in a distal direction, causing stresses in the mesial region by a leverage
effect of the implant body. This interesting effect was not found in long monocortical anchorage, including anchorage in the cancellous bone. Apparently peri-implant stresses caused by this leverage effect can be prevented by choosing longer mini-implants.

**Bicortical anchorage**

The cervical bone stress was clearly lower in the bicortical variant than in all variants of monocortical anchorage. The reduction factor was 2–3, leading to low stresses of the cervical bone in the direction of the applied force. In contrast to monocortical anchorage, the mesial region of the peri-implant bone did not show relevant stresses. Obviously, in bicortical anchorage the mini-implant is completely fixed by the opposite cortical bone, effectively preventing a leverage effect of the implant body. The overall stresses of the cervical bone are very low in bicortical anchorage; thus the risk of loosening or loss of the mini-implant is reduced in comparison to monocortical anchorage.

**Conclusion**

From a biomechanical point of view bicortical mini-implant anchorage is more favourable than monocortical anchorage, because the induced stress in the peri-implant bone is less. Therefore, bicortical anchorage should be especially considered in challenging clinical situations requiring heavy anchorage. This implies a beneficial morphology, allowing a bicortical insertion without increased risk of injury to the adjacent structures.

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