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Three-dimensional numerical simulation of dental implants as orthodontic anchorage

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SUMMARY Endosseous oral implants have been used as orthodontic anchorage in subjects with multiple tooth agenesis, and their application under orthodontic loading has been demonstrated clinically and experimentally. The aim of this investigation was to examine three-dimensional (3D) bone and implant finite element (FE) models. The first model assumed that there was no osseointegration and the second that full osseointegration had occurred. These models were used to determine the pattern and distribution of stresses within the ITI-Bonefit® endosseous implant and its supporting tissues when used as an orthodontic anchorage unit. The study examined a threaded implant placed in an edentulous segment of a human mandible with cortical and cancellous bone.

The results, using both models, indicated that the maximum stresses were always located around the neck of the implant, in the marginal bone. Thus, this area should be preserved clinically in order to maintain the bone–implant interface structurally and functionally.

Introduction

The number of adults seeking orthodontic treatment has increased significantly in recent decades (Melsen *et al.*, 1987; Kalia and Melsen, 2001). These patients often require multidisciplinary treatment, especially when they are restoratively or periodontally compromised, or have multiple tooth agenesis.

The successful use of osseointegrated implants as anchorage elements for orthodontic force application has been reported in several experimental studies (Smalley *et al.*, 1988; Turley *et al.*, 1988; Linder-Aronson *et al.*, 1990; Wehrbein and Diedrich, 1993; Wehrbein *et al.*, 1997; Majzoub *et al.*, 1999) and clinical case reports (Roberts *et al.*, 1989, 1990; Haanaes *et al.*, 1991; Higuchi and Slack, 1991; Ödman *et al.*, 1994; Wehrbein *et al.*, 1996). For orthodontic applications, implants have been inserted into a number of areas, including the retromolar region (Roberts *et al.*, 1990), the alveolar process (Turley *et al.*, 1988; Ödman *et al.*, 1988, 1994; Haanaes *et al.*, 1991) and also the midpalatal suture (Fontenelle, 1991; Wehrbein *et al.*, 1996).

The clinical success of an implant is largely determined by the manner in which the mechanical stresses are transferred from the implant to the surrounding bone without generating forces of a magnitude that would jeopardize the longevity of the implant and prosthesis (Skalak, 1983). Functional forces, biomechanical characteristics, and stress transfer to the surrounding tissues are among the factors involved in the design of dental implants. However, these data are currently unavailable.

The finite element (FE) method has been used in specific dental applications, including the analysis of stress around osseointegrated implants (Williams *et al.*, 1990; Clelland *et al.*, 1991).

The objectives of this three-dimensional (3D) FE study were to investigate orthodontic loading simulations on a single ITI-Bonefit® endosseous implant (Institut Straumann A.G., Waldenburg, Switzerland) and its surrounding osseous structure, to analyse the resultant stresses, and to identify the changes in the bone adjacent to the implant following orthodontic loading.

Materials and methods

The implant and its mechanical properties

The models, examining the implant with and without osseointegration, consisted of the endosseous dental implant and the surrounding bone. The geometric design (and mesh) of the implant is shown in Figure 1. The implant (diameter 4.1 mm, length 10 mm) is a threaded endosseous implant made of commercially pure titanium.

In the first model, the contact between the implant and the bone was simulated, while in the second it was assumed to be 100 per cent osseointegrated. The boundary conditions were defined to simulate how the model was constrained and to prevent it from free bodily movement. Thus, it was assumed that the lower part of bone which simulates the mandible was fixed.

Finally, both the surrounding bone and the implant were modelled with a linearly elastic behaviour, and the mechanical properties of Young's modulus (E) and

Table 1 Elastic material properties used in the finite element model.

Materials	Young's modulus ($1\ \text{MPa} = 10^6\ \text{Pa}$)	Poisson's ratio
Compact bone	13 760 MPa	0.30
Cancellous bone	7930 MPa	0.30
Titanium	110 000 MPa	0.35

Poisson's ratio (ν) included. The values of *E* and ν (Table 1) that were used in the simulations for cortical bone, trabecular bone, and titanium were obtained from the literature (Carter, 1978; Carter *et al.*, 1980). Isotropic and homogeneous elastic properties for bone were assumed. The osseous model was composed of cancellous bone surrounded by cortical bone.

Implant loading

Simulated horizontal loads of 20 N, at 90 degrees from the long axis, were applied to the top of the implant. The study simulated loads in a horizontal direction, similar to a distal–mesial orthodontic movement.

Stresses (in MPa = 10^6 Pa) were calculated and presented as coloured contour bands; different colours representing different stress levels in the deformed state. Positive or negative values of the stress spectrum indicate tension and compression, respectively.

FE model

The FE method is a theoretical technique used in engineering. For FE method analysis, a structure is modelled with discrete-element mathematical representation by subdividing it into simpler geometric shapes, or elements, whose apices meet to form nodes. The elastic constants, *E* and ν, are specified for the materials modelled, with each element retaining the mechanical characteristics of the original structure.

Stresses and displacements in the implant and the bone can be calculated by the FE method. Generally, this method results in a set of equations which can be solved. The solution provided by this program represents the displacements of all nodes of the FE mesh simulating the complex bone–implant. A code was then developed to determine the elastic stress by means of the 3D von Mises stress norm.

The whole mesh (Figure 1) had 2492 nodes and 10 213 elements (tetrahedra), and a code developed for solving the discrete variation inequality. The Modulef program (Inria, Paris, France) was used for the 3D model. It was operated on an IBM RISC 6000 Unix Workstation and a typical run took less than 1 hour.

In the first FE model the materials were elastic and it was assumed there was no osseointegration. Thus, a

Figure 1 Schematic three-dimensional image of the ITI-Bonefit® implant.

contact condition needed to be imposed with no penetration of one material into the other and with friction ignored. This corresponded to the classic frictionless contact between two elastic bodies (Kikuchi and Oden, 1988). The FE method, in this model, led to a discrete variational inequality that was solved using a penalty–duality algorithm (Viaño, 1986; Burguera and Viaño, 1995). This model configuration represented the situation immediately after implantation when the implant was totally surrounded by cancellous bone (Figure 2a).

In the second model, as it was assumed that the material was elastic and that osseointegration was complete, the classic FE model in elasticity could be considered. After osseointegration (Figure 3a), there was no difference at the contact boundary between the surrounding bone and the implant. Boundary nodes of both parts were designed to be common and so it could be assumed that the complex bone–implant was a unique domain composed of two mechanical parts (each of them with different elastic coefficients); the surrounding bone and implant.

Results

The stress concentration at the *y*-plane on the bone surrounding the dental implant around the neck of the implant is depicted in Figure 2a (model 1). Von Mises norm is a scaler representation of the general effective stress in a material. In these images, the scale for stress runs from 0 MPa (blue) to the highest stress values (red). In model 1, the stress was mainly concentrated at

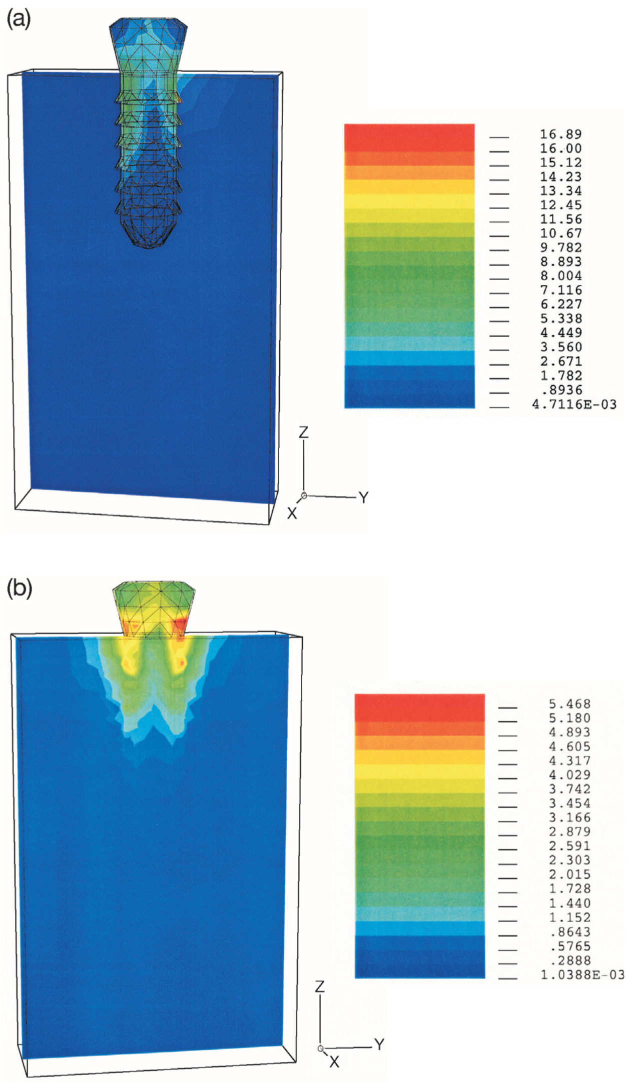

Figure 2 Von Mises stress fields distribution following the application of a 20 N horizontal force in (a) model 1, a non-osseointegrated bone–implant complex and (b) model 2, a fully osseointegrated bone–implant complex. The colour scale indicates the magnitude of the stresses.

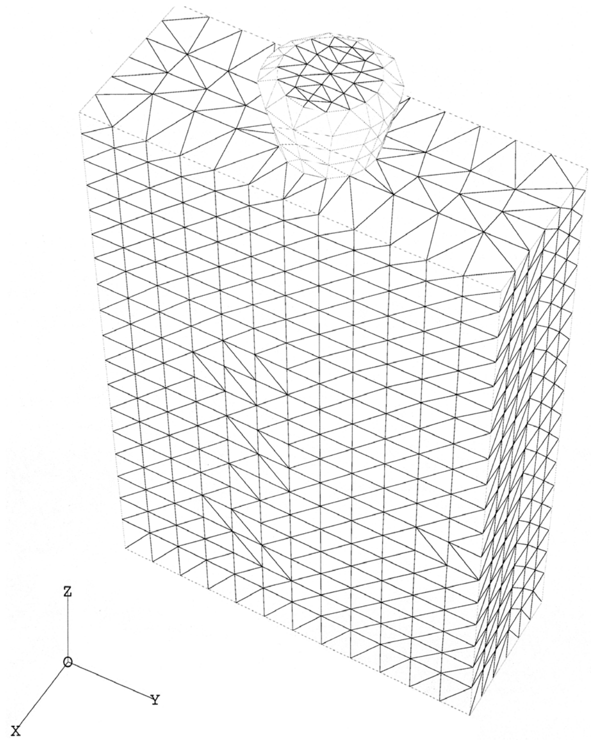

Figure 3 A three-dimensional geometric element model of the implant and the surrounding bone.

the neck of the implant and at the closest surrounding bone (green areas). Figure 2b shows the stress concentration at the *y*-plane on the bone surrounding the dental implant around the neck of the implant after complete osseointegration (model 2). This stress was chiefly concentrated (yellow–red areas) at the neck of the implant and at the level of the cortical superficial bone. The stresses decreased in the cancellous bone area. The dental implant behaved as a rigid structure, with the highest stresses concentrated at the first cervical thread, decreasing uniformly to the apex. Analysis of the *x*-plane at the cortical bone around the implant revealed that the cervical margin and the bone around the first thread of the dental implant were the most stressed areas. The stress distribution at the mesial and distal sides showed almost symmetrical behaviour but *vice versa*; the maximum compressive stress was localized mesially and the maximum tensile stress distally. If both models are compared, it can be observed that the stresses were less and more evenly distributed in model 1 (initial stability) than in model 2 when osseointegration was assumed.

Discussion

Orthodontic anchorage provided by teeth and/or intraoral structures (e.g. dental implants) and movement of teeth in the posterior direction are of fundamental importance in orthodontic treatment. The aim of the present study was to analyse the deformation of the bone surrounding a dental implant in response to an orthodontic load. In this investigation, the loads were analysed in the horizontal plane, because simulated and orthodontic loadings in sliding mechanics were applied. This method assumes that the implant is intimately in contact with bone, thereby simulating osseointegration or immediate stability clinically after implantation. These two situations are clinically relevant to improve the understanding of biomechanical aspects of dental implants. However, this is still an imperfect approximation of the clinical situation because of the assumptions made for the models (homogeneity of the materials, perfect bonding between the bone and implant, etc.). These assumptions have to be considered when interpreting the results (Geng *et al.*, 2001).

Regarding the first assumption, the models did not include the heterogeneous aspects of bone, because its real structure cannot be modelled. The model was created taking into account the general distribution and different characteristics of cortical bone and cancellous bone. In the absence of further information concerning the precise material properties of bone, and in line with the majority of FE studies, it was assumed that cortical and cancellous bone were isotropic, homogeneous, and linearly elastic (Meijer *et al.*, 1995, 1996). Regarding the second assumption, more detailed models have been under development to study the effects of bonding on the stress around implants, as 100 per cent fixation (total osseointegration) may not be possible to achieve in clinical practice (it is an ideal and unrealistic assumption) and because 100 per cent bone apposition is not always obtained at the surface of the endosseous dental implant. The formation of bone at the implant–tissue interface is essential in achieving rigid osseous fixation of the implant and has been considered as an indication of success. However, sufficient data concerning osseous healing and the interface between dental implants and bone are still unavailable. The first model is of interest as oral surgeons sometimes use immediate loading of implants. The lack of initial post-operative implant stability is recognized as an important factor in the loosening/failure process of implants. The stress distribution in model 1 demonstrated initial stability because the dental implant did not suffer from a failure to osseointegrate.

Analysis of the implant after horizontal loading showed that the cervical third of the mandibular bone supported important loadings both in terms of stress magnitude and distribution. The highest stress concentration in the implant was localized at the cervical margin and at the first threads of the implant. Similar results have been reported (Clelland *et al.*, 1991; Meijer *et al.*, 1995; Barbier *et al.*, 1998; Vásquez *et al.*, 2001). In agreement with Vásquez *et al.* (2001) it was found that these stresses were of such low magnitude that they were unable to produce permanent failure of the implant. However, using the FE method, several authors (Borchers and Reichart, 1983; Meijer *et al.*, 1995; Barbier *et al.*, 1998, Chen *et al.*, 1999) have found that the highest risk of bone resorption occurs at the neck region of an implant. It was found that stress distribution was less concentrated and more uniformly distributed at the neck region of the first (initial stability) than of the second (osseointegration) model because of a different biological adaptation to loads (bone elasticity versus formation osseous union).

It is important to note that osseointegrated implants are able to support orthodontic loading and may function as adequate anchorage units. In most clinical cases, this anchorage unit will subsequently be used for restorative purposes. It is very important, therefore, not to jeopardize the bone–implant interface with traumatic loading situations. The results of the present study illustrate that there is a greater risk of overload at the mesial and distal bone. This should be taken into account in patients where a narrow alveolar bone ridge exists, as in some adult patients with several missing posterior teeth where an endosseous implant is being used for orthodontic anchorage. The implant is often placed in areas with local defects (e.g. dehiscences, recent dental extraction or a narrow alveolar bone ridge). In these cases, it may be advisable to consider osseous regeneration techniques prior to implantation. Because orthodontic loading does not necessarily mean that the ultimate strength of bone tissue will be exceeded, continuous loading is more likely to cause fatigue damage (bone microcracks, marginal bone resorption) that could jeopardize the anchorage unit. From a mechanical point of view, the presence of bone defects seems unfavourable due to the lack of bone support. Conversely, peri-implant bone stresses and strains are not only a function of the *in vivo* loading conditions, but are also determined by the bone quality (bone mechanical properties) and quantity (cortical bone thickness, cancellous bone density), periodontal status, oral hygiene and numerous other factors that may play a role in marginal bone remodelling (Carmagnola *et al.*, 1999; Van Oosterwyck *et al.*, 2002).

Conclusions

The assumptions established in the construction of the 3D FE model showed the area with the highest stress to be around the dental implant when used as orthodontic anchorage and the surrounding bone was the cervical margin. This finding is clinically important in order to preserve the bone–implant interface in this area. Therefore, when osseointegrated implants are primarily used as anchorage for orthodontic purposes and then as fixed prostheses, the functional and structural union of titanium to bone should be preserved. A lack of bony support for the implant (pre- or post-implantation) represents an unfavourable situation from a biomechanical point of view that should be considered and solved. As clinical problems mostly occur at the marginal bone region (bacterial plaque accumulation, over-contoured abutments, infections, osseous defects), attention should be focused on this region.

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