

Three-dimensional Finite Element Stress Analysis of a Cuneiform-Geometry Implant

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Purpose: The biomechanical behavior of an osseointegrated dental implant plays an important role in its functional longevity inside the bone. Studies of this aspect of dental implants by the finite element method are ongoing. In the present study, a cuneiform-geometry implant was considered with a 3-dimensional model that had a mesh that was finer than in the models commonly found in the literature. **Materials and Methods:** A mechanical model of an edentulous mandible was generated from computerized tomography, with the implant placed in the left first premolar region. A 100-N axial load was applied at the implant abutment, and the mandibular boundary conditions were modeled considering the real geometry of its muscle supporting system. The cortical and trabecular bone was assumed to be homogeneous, isotropic, and linearly elastic. **Results:** The stress analysis provided results that were used to plot global and detailed graphics of normal maximum (S1), minimum (S3), and von Mises stress fields. The results obtained were analyzed and compared qualitatively with the literature. **Discussion:** Quantitative comparisons were not performed because of basic differences between the model adopted here and those used by other authors. The stress distribution pattern for the studied geometry was similar to those found in the current literature, but insignificant apical stress concentration occurred. The stress concentration occurred at the neck of the implant, ie, in the cortical bone, which was similar to results for other implant shapes reported in the literature. **Conclusion:** The studied geometry showed a smooth stress pattern, with stress concentrated in the cervical region. The values, however, were within the range of values found in the cortical layer far from the implant, caused by the muscular action. No significant stress concentration was found in the apical area. (More than 50 references.) INT J ORAL MAXILLOFAC IMPLANTS 2003;18:675–684

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Dental implants are frequently submitted to multidirectional loads originating in the stomatognathic system,^{1–5} and the bone stress distribution pattern is highly relevant to the bone-implant relationship and consequently to its longevity.^{3–7} The biomechanical behavior is directly dependent not only on the implant geometry, but on the whole design of the implant, which includes its shape and material, as well as the prosthesis it supports.^{8,9} Considering the similarity of prostheses and materials in all prosthetic solutions, valuable contributions may come through the study of the influence of implant shape.⁸ Mechanical analysis using the finite element method (FEM) has been employed by many authors to understand the biomechanical behavior around dental implants^{3,6,10–37} with a suitable degree of reliability and accuracy, but without the risk and expense of implantation.³

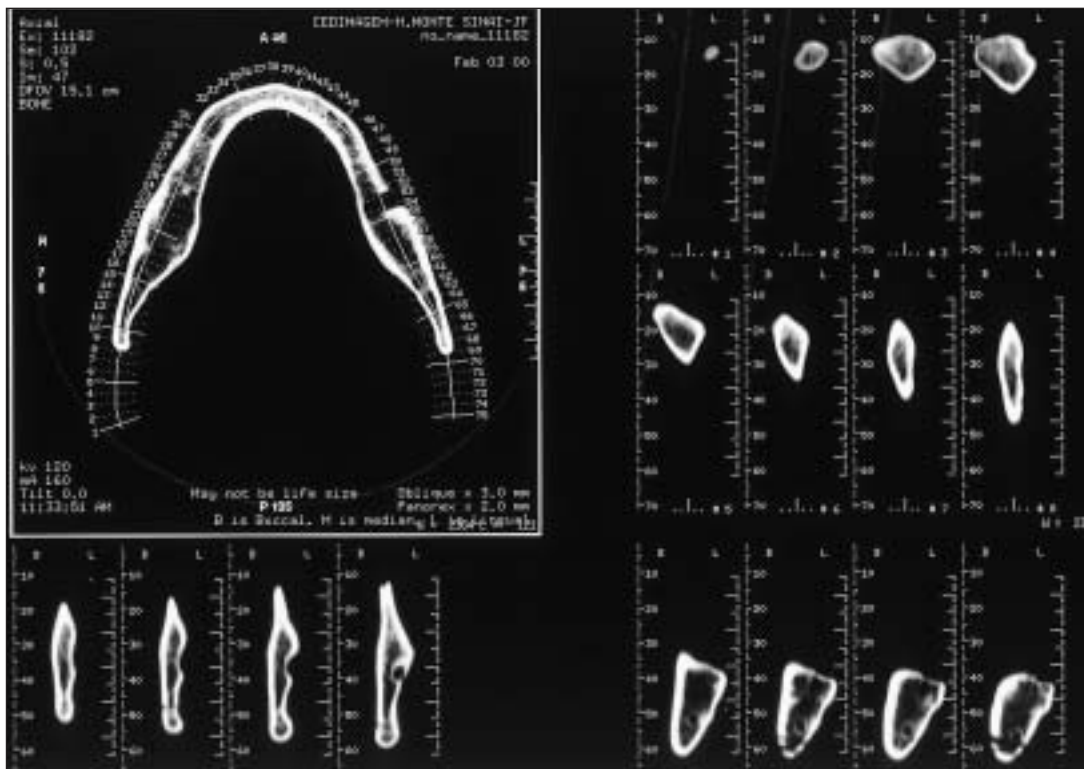


Fig 1 To begin creating the model used for FEM in this study, CT of the mandible was performed.

The degree of accuracy of the FEM is related to knowledge of the real load and supporting conditions.³⁸⁻⁴³ Different studies agree that biomechanical behavior plays an important role in the survival of an implant,^{3-7,10,28-30,44} that geometry is a key factor associated with stress distribution,^{5,13,45,46} and that the FEM can be a reliable method for studying the biomechanical behavior of implants.^{3,5-7,14,15,26,28,37,39} Previous attempts to model the tissue-implant interaction have, for the most part, been limited to the use of 2-dimensional (2D) analysis,^{5,7,10,13} and more refined models that bear a closer resemblance to actual anatomic and physiologic structures can make the method's results more reliable.^{38,39,43} The influence of environmental conditions around the implant, such as different degrees of bone-implant contact,^{6,11} presence of cortical or cancellous bone,^{21,23} and mandibular body deformation,⁴⁷ have also been studied and reproduced.^{3,6,14,15,28,44}

The osseointegrated implant interface is rigid and transmits occlusal loads directly to adjacent bone. This condition can produce high levels of stress,⁴⁸ concentrated mainly in the neck and apex region,⁴⁹⁻⁵² that can affect the bone physiology.^{53,54} Many authors have studied and tried to develop different ways to compensate this situation through the use of devices such as an intramobile element^{29,55,56} or a resilient collar around the implant neck⁵⁷ and

even with cementum/periodontal ligament formation around the implant.⁵⁸⁻⁶⁰ Different geometries have been studied, and attempts have also been made toward shape optimization that enhances the relationship between bone and implant, allowing for improved biomechanical performance.^{27,29,30}

The goal of this work was to analyze the stress distribution around a cuneiform implant using accurate modeling capable of obtaining more precise data, thereby enhancing the results found in the current literature.

MATERIALS AND METHODS

The starting point of this study was the development of an accurate model of an edentulous mandible, which was essential for obtaining more precise results.^{38,39,43}

Geometric Modeling

Initially, computerized tomography (CT) (Tomograph Pro-Speed; GE Medical Systems, Fairfield, CT) of an actual human mandible was obtained according to the description of Inou and coworkers⁴³ (Fig 1). With the help of a scanner, the images obtained were converted into digital data and transferred to a CAD program (AutoCAD; Autodesk, San

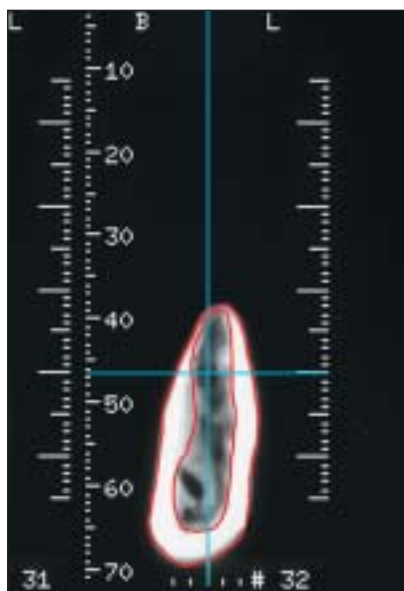


Fig 2a Transverse section #32, where the implant was placed.

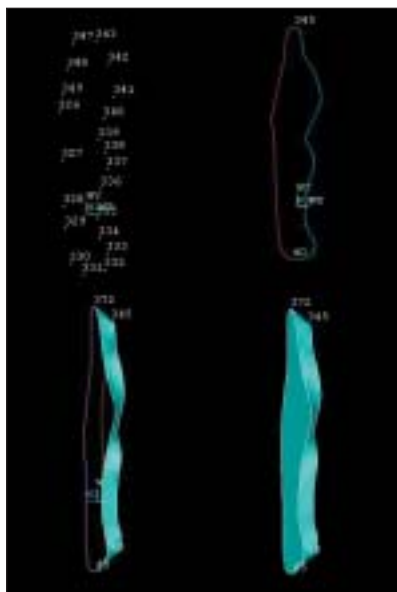


Fig 2b Assembly of the geometric model, points, lines, surfaces, and volumes.



Fig 2c Computational geometric model of the mandible.

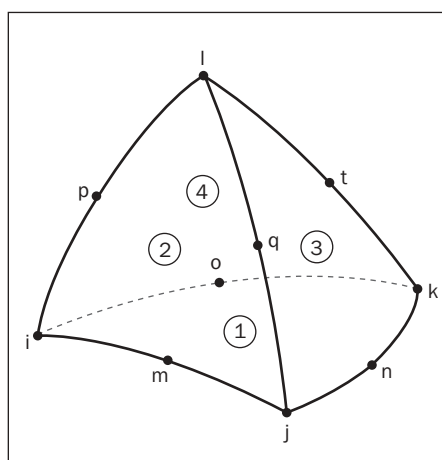


Fig 3 (Left) Tetrahedral isoparametric square element.

Fig 4 (Below) Mathematical model of the mandible showing the finite element mesh.



Rafael, CA), where the coordinates of the contouring points were extracted from these plots and joined to form partial volumes that together defined the final geometry (Figs 2a and 2b). This sequence was done on one side and repeated to obtain the other side. Through this process the CT data were converted into a 3D solid model (Fig 2c).

Mathematical Modeling

As is routine when using FEM, the geometric model was meshed; in the present case, this was done with tetrahedral isoparametric quadratic elements, utilizing 4 triangular faces, 4 vertices, and 10 nodes (Fig 3). The displacement of these nodes was found and used in the calculation of the stress distribution inside the structure. The grid reached 85,800 elements with 362,610 degrees of freedom.

Of this total number of elements, 67,120 of them, which corresponded to 276,960 degrees of freedom, were distributed in the region of the left premolar, between sections #30 and #34 where the implant was placed (Fig 2a), since the greatest numeric accuracy was desired in this area. The remaining elements were distributed throughout the mandibular body (Fig 4). The Bioform implant (Maxtron, Juiz de Fora, MG, Brazil) was modeled by a CAD program (Fig 5a) together with its prosthetic abutment in a fine mesh (Fig 5b) with 17,193 elements and 80,134 degrees of freedom (Table 1).

Load and Supporting System

For the boundary condition of the model, a supporting system was set up. The model was supported by the muscles of mastication³⁵ and the

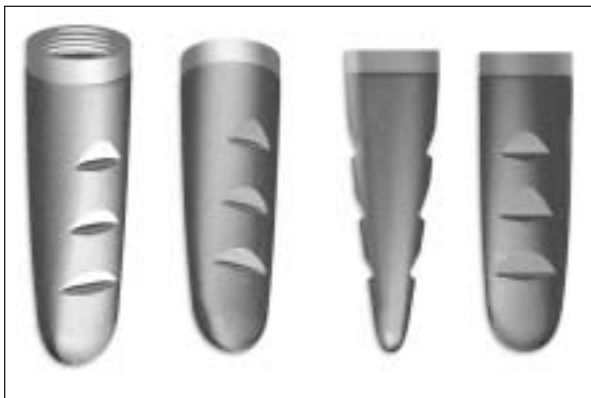


Fig 5a Views of the Bioform implant with its cuneiform geometry. It has 3 notches on each major side of the body to enhance mechanical retention.



Fig 5b Mathematical model of the Bioform implant 413 and the abutment.

Table 1 Mesh Data			
Region	Elements	Nodes	Degrees of freedom
Implant/abutment	17,193	26,708	80,134
Sections 30 to 34 + implant/abutment	67,120	92,320	276,960
Complete model	85,800	120,870	362,610

Table 2 Distance Vector Components (mm)			
Vector distance	X direction	Y direction	Z direction
r_M	0.0	28.07	33.01
r_T	0.0	30.61	5.27
r_{PL}	0.0	9.56	6.31
r_{PM}	0.0	27.67	38.97
r_{P_0}	0.0	80.63	23.89

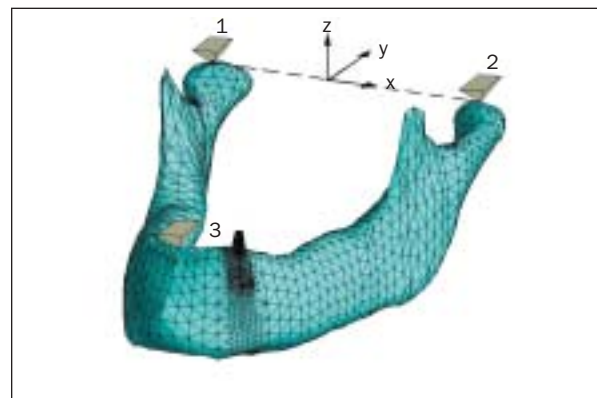


Fig 6 Model with restraining adopted. Point 1 = Translation restrained in directions x, y, z; point 2 = translation restrained in directions y, z; point 3 = check point.

temporomandibular joints.⁶¹ The forces generated by the muscles of mastication (temporalis [T], masseter [M], medial pterygoid [PM], and lateral pterygoid [PL]) were calculated, based on their transverse sections, according to Inou and coworkers.⁴³ Adapting the data from this reference, the relationships between the muscle actions were as follows:

$$M = 1.72 PL \quad (\text{equation 1})$$

$$T = 0.99 PL \quad (\text{equation 2})$$

$$PM = 1.15 PL \quad (\text{equation 3})$$

The model was restrained as shown in Fig 6, and an axial load $P_0 = 100$ N was applied at the top of the abutment. The values of the muscular forces to keep the model equilibrated were obtained from the following equation.

$$2M \times r_M + 2PM \times r_{PM} + 2PL \times r_{PL} + 2T \times r_T + 100\mathbf{u} \times r_P = 0 \quad (\text{equation 4})$$

where r_M , r_{PM} , r_{PL} , r_T , and r_P are the distance vectors from the load application points of the M, PM, PL, T, and P_0 (axial load over the implant) to the x (1-2) axis, that pass across the center of the condyles, respectively (Fig 6). In this equation, the symbol \mathbf{u} indicates the unitary vector in the implant axis direction and the symbol \times denotes the vector product.

The positions of the vectors of the muscular forces and the axial load are indicated in Table 2.

The muscle positioning on the mandibular body (Fig 7a) was approximated based on descriptions found in the literature,⁶²⁻⁶⁶ and the results of the muscular forces were considered to be acting on the centroid of the nodes of the elements that define the muscular action areas⁶⁷ (Fig 7b).

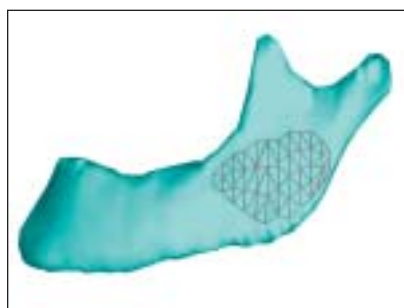


Fig 7a Outline of distribution of the masseter muscle.

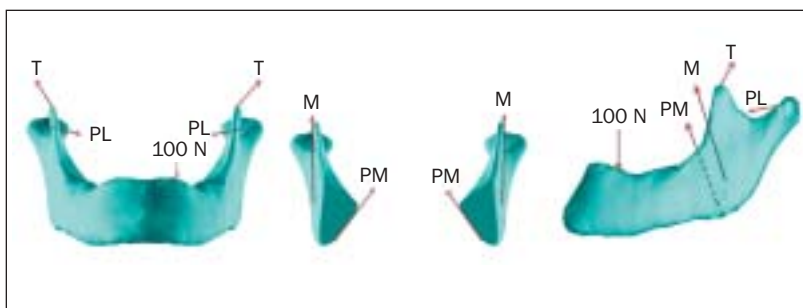


Fig 7b Directions of the applied muscular forces.

Table 3 Directional Cosines of the Resultant Muscular Forces (Right Side)

Muscle	Cos (α)	Cos (β)	Cos (γ)
Masseter	-0.043	-0.011	0.999
Medial pterygoid	0.587	-0.165	0.792
Lateral pterygoid	0.714	-0.692	0.106
Temporalis	-0.325	0.219	0.920

Table 4 Elasticity of Materials Used

Material	Modulus elasticity	Poisson's ratio	References
Cortical bone	13,700 MPa	0.30	3,6,11,16,21-25, 27,28,44,46,50
Cancellous bone	1,370 MPa	0.30	11,16,21,23-25, 28,44,46
Titanium	110,000 MPa	0.33	16,22,25,26

The directions of the forces were established by the cosine extracted from the geometry considered (see Table 3).

Equation 4 and the relationships described led to the resultant values, which were: M = 59.23 N; PM = 39.60 N; PL = 34.44 N; and T = 34.09 N.

Material Properties

In the absence of information about the bone's precise material properties, assumptions were made according to the majority of studies that used FEM (Table 4).

The areas of cortical and cancellous bone were assumed as defined by the CT sections in the mandibular body (Fig 2), and the 2 types of bone modeled (cortical and cancellous) were considered isotropic, homogeneous, and linearly elastic.

Implant System, Load Positioning, and Interface Conditions

The cuneiform implant used here is 13 mm long and 4 mm in diameter (Figs 5a and 5b). This geometry has an interesting resemblance to natural roots^{68,69} and also has a high degree of applicability because of its bone-induced expansion capacity.⁷⁰ Its design also presents 3 notches on each side of the body, which provides better mechanical retention during prosthetic management. The dimensions were chosen according to the majority of FEM studies. It was assumed that the implant was placed between sections #30 and #34, ie, the premolar region, which is

representative of the average force acting in the mouth.^{8,39} A vertical load of 100 N was applied on top of the abutment in the direction of the long axis of the implant.^{17,28,32,44} A layer of cortical bone 2 mm thick was contoured around the implant neck, and the body of the implant was embedded in the cancellous bone. A fixed bond, ie, total osseointegration, between bone and implant along the whole interface was assumed, which meant that under the applied load on the implant, relative motion between bone and implant did not occur.

Operational Conditions

The analyses were accomplished with the Ansys software program (Ansys Corporate, Canonsburg, PA) and processed by a personal computer (IBM, White Plains, NY).

RESULTS

The stress analysis executed by Ansys provided results that enabled the tracing of global and detailed graphics of the maximum (S1) and minimum (S3) principal stresses and the von Mises stress field. Stress contours were color-coded and explained for each figure. All stress values were indicated in mega pascals (MPa).

Figures 8a to 8d display the global results as the mandibular body deformation presented a medial convergence⁴⁷ (Fig 8a). The maximum and minimum

Fig 8 Global results.

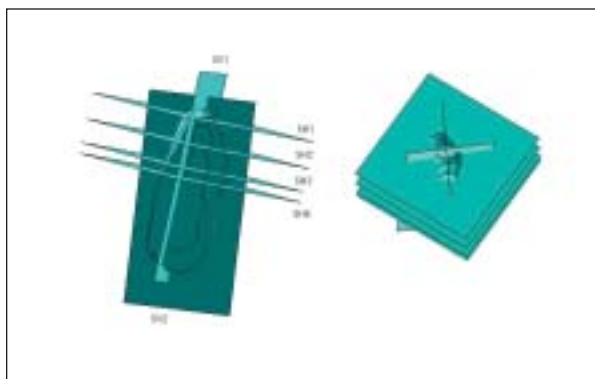
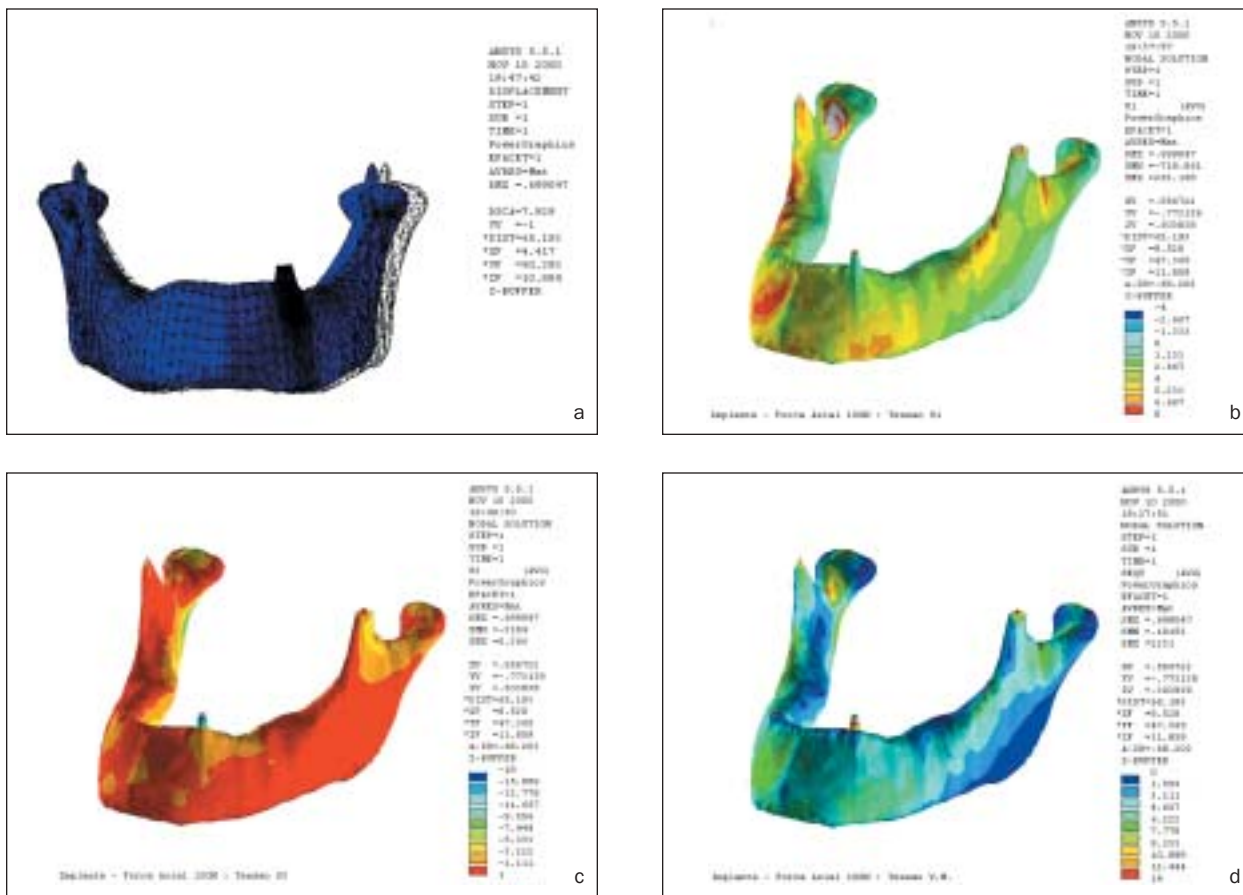


Fig 9 Positioning of the horizontal and vertical sections.

principal stresses showed smooth distribution, with stress concentration only at the muscular insertions (Figs 8b and 8c); the same occurred with the von Mises criteria stresses (Fig 8d).

For enhanced accuracy, analysis graphics were made of the region between the sections #30 and #40

to show the values of principal maximum, minimum, and von Mises criteria stresses. The images were presented in 4 transverse and 2 longitudinal sections (Fig 9) and 2 more detailed sections from the implant's neck region (Figs 10a to 10c). The transverse sections were obtained as follows: SH4 is tangent to the apex of the implant, SH3 is 2 mm from SH4, SH2 is 4.5 mm from SH3, and SH1 is 5 mm from SH2.

All graphics showed a smooth distribution of stresses, with no significant concentration at the apex. An area of stress concentration was present on one side of the neck. However, the values were equivalent to those registered at the insertion of the masseter. The highest stress concentration occurred at the superior side of the cortical layer, but the values shown within the figures were within the same order of those encountered in the cortical layer under the masseter. Low areas of stress were observed on the upper side of the re-entrances (Figs 10a to 10c) present in the implant body but were within the bony physiologic limits. The cross sections showed stresses along the implant from the top to the apex.

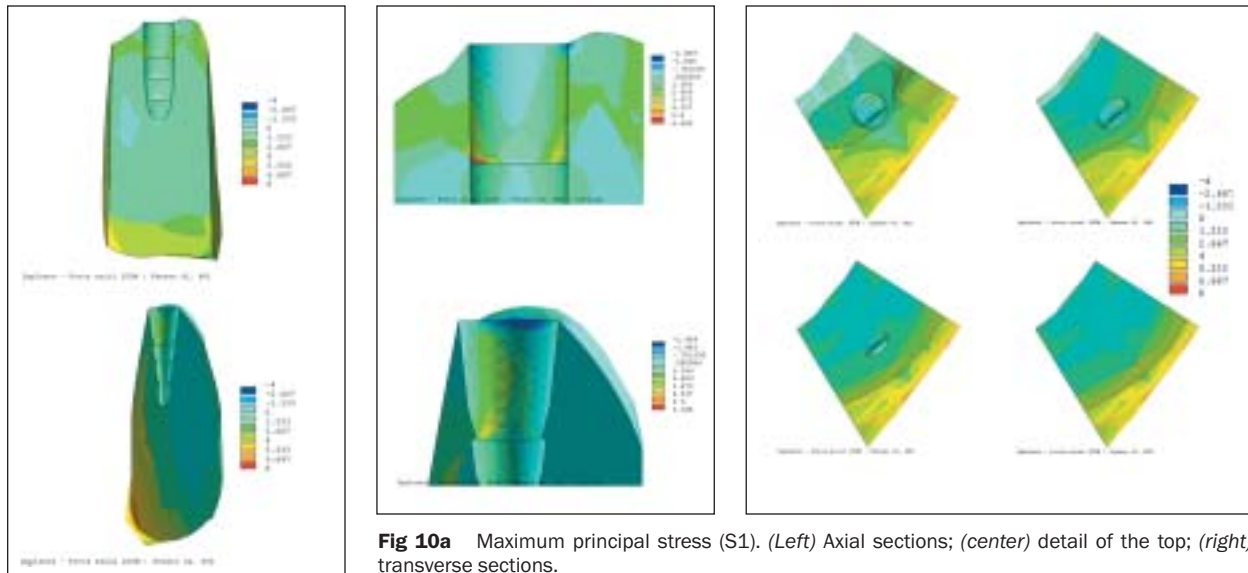


Fig 10a Maximum principal stress (S1). (Left) Axial sections; (center) detail of the top; (right) transverse sections.

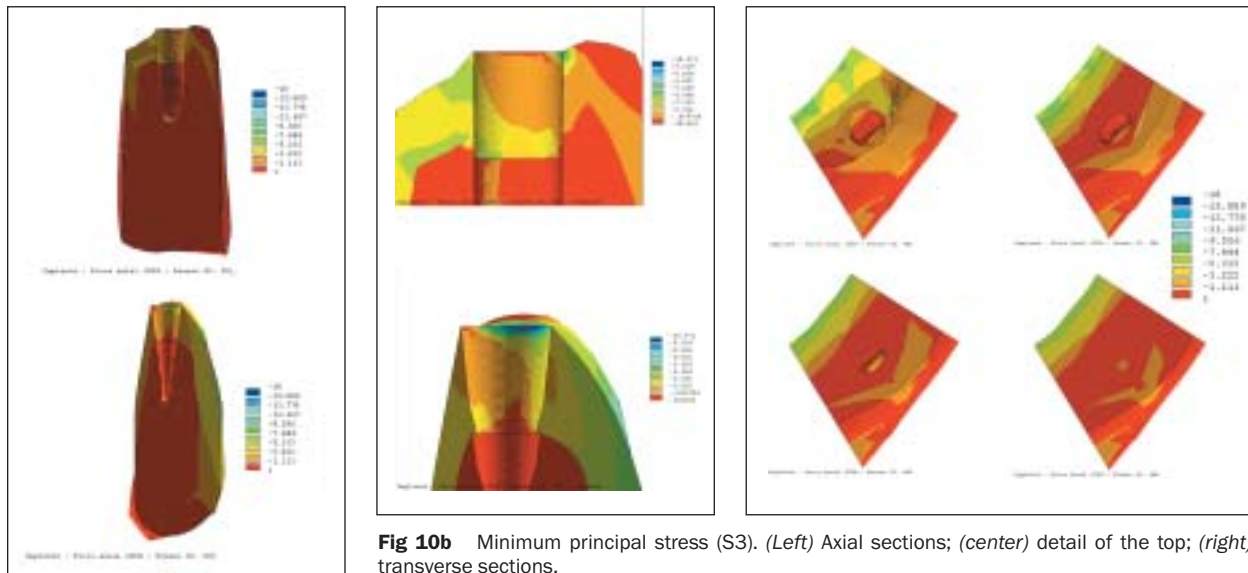


Fig 10b Minimum principal stress (S3). (Left) Axial sections; (center) detail of the top; (right) transverse sections.

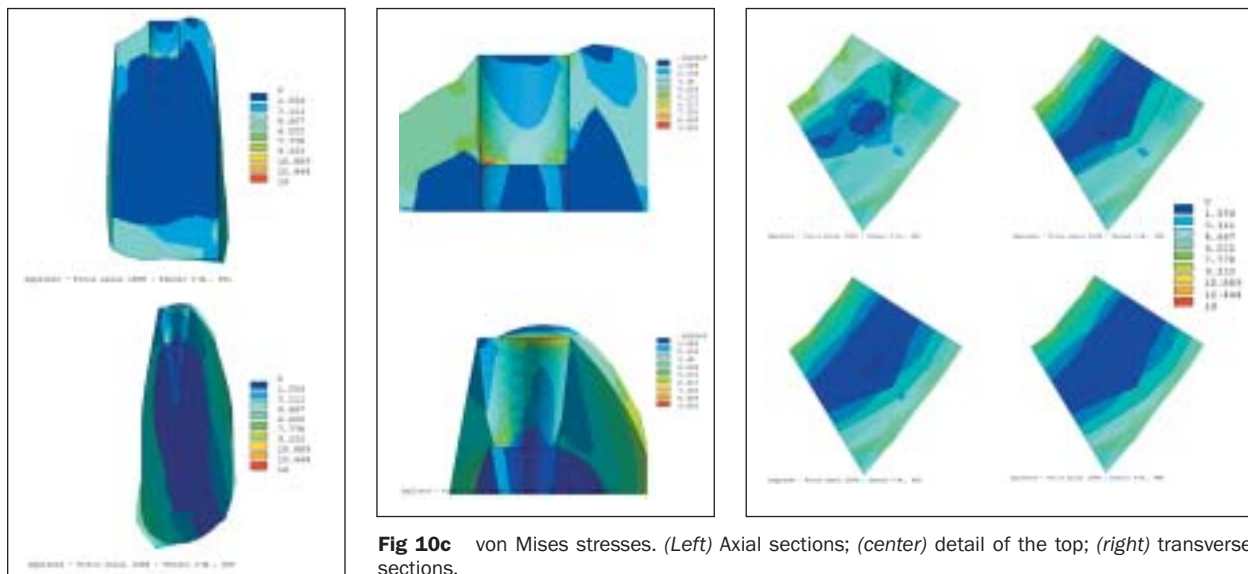


Fig 10c von Mises stresses. (Left) Axial sections; (center) detail of the top; (right) transverse sections.

DISCUSSION

In this study, the global analysis showed deformation of the mandible similar to that seen in many reports in the literature.^{38-40,47,66} The importance of this deformation in the results of FEM studies has been noted by some authors.^{38,39,42} Three-dimensional modeling^{6,38,39,43} utilizing a fine mesh with a great number of elements³⁸ contributes to the reliability of FEM.

The modeled muscular force action at the bone surface generated stresses as high as those obtained around the implant, as shown in the results. This fact provides a qualitative means of comparing the stress levels achieved and suggests that modeling of the entire mandible is important and cannot be neglected. Models with supporting systems that do not consider these factors^{11,13,28} can provide unrealistic results. A comparison of different modeling conditions can serve as a reference but does not have conclusive value. In the area around the implant, the results of the present analysis showed stress concentration in the cortical layer facing the implant neck, similar to the majority of previously reported results.^{5,7,14,15,23,49,54}

A comparison of the data obtained from this study to those obtained from 3 works presented by Rieger and coworkers,^{7,14,15} who studied different geometries modeled under the same conditions, shows that the stress concentration at the implant neck of the cuneiform-shaped implant was more favorable than that found at cylindrical and tapered threaded implants. Bone loss in this area has been correlated to this stress concentration, and some studies have used these values, together with clinical data, to confirm this statement.^{5,7,10,11,17,44,46,50,52,59} In this study, the stress concentration occurred only on one side and not around the neck as previously related.^{14,15,17,46,51,52} Rieger and coworkers¹⁵ also suggested that the cylindrical geometry analyzed in their survey, with the exception of the maximum stress found near the neck and the apex, transfers relatively low stresses to the bone along the body of the implant. This situation can lead to pathologic bone loss near the middle of the implant as a result of atrophy and at the extremities as a result of the excessive stress. The geometric contour of the implant studied demonstrated a smooth distribution of stresses and induced a gradual distribution of the load from the top to the apex region. Also, results related to other geometries have frequently described an apex stress concentration.^{5,13-15,27,34,44} In the present work, stress concentration at the apex of the implant was insignificant and probably should not be considered.

Siegele and Soltész¹³ described stress concentration at the apex of a cuneiform implant and related this to the small area at the apex of its geometry versus cylindrical implant geometry. However, the results were obtained with a nonrigid interface, ie, a nonosseointegrated implant model. Rieger and associates¹⁴ stated that tapered implants are better than cylindrical implants at avoiding punching stresses, and Rieger and coworkers¹⁵ and Sodré³⁴ related that conical geometry had better biomechanical performance. Adams¹² illustrated that a cylindrical implant design would direct most of an applied axial load to the apical region and recommended tapered geometries for better stress distribution. Deines and colleagues⁴⁵ described better performance for natural tooth geometry (ie, cuneiform) compared with other standard implant geometry.

A geometry that takes stress away from the bone crest should be chosen for clinical use, as affirmed by Akpinar and coworkers.⁴⁶ This did not totally occur with the present geometry, but the stress distribution pattern of this analysis showed values of the same magnitude at the neck and at the muscle insertion. As stated by Rieger and associates,¹⁴ low stresses can be as problematic as high stresses. In this geometry, in some areas where the stresses were very low, such as the notches in the implant body (Figs 10a to 10c), the stress values were within the limits that are sufficient to maintain normal bone physiology.^{5,7,11,53}

Because of the differences among the models used by other authors, a quantitative comparison between the different implant designs is not reasonable. However, a qualitative analysis of the presented results demonstrated that the cuneiform geometry exhibited acceptable biomechanical performance.

CONCLUSIONS

Despite the limitations of the methodology that considered the bone homogeneous, the results of static load and linear analysis support the following conclusions:

- The cuneiform-geometry implant conforms to the pattern related in the literature, with a stress concentration in the cortical region. However, the values were within the same range of the stress level in the cortex when subjected to muscular action.
- No considerable apical stress concentrations were found.
- The cuneiform geometry appeared to distribute the stress in a smooth pattern.
- The muscular action caused areas of stress concentration, suggesting that modeling of musculature is very important to the accuracy of the data obtained.

- Under muscular action, the mandible deforms in such a way that the condyles converge medially.
- The modeling methodology, conditions of the support and load system, and the finest anatomic and functional variations played important roles in the results.

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