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STRESS ANALYSIS WITH NONLINEAR MODELLING OF THE LOAD TRANSFER CHARACTERISTICS ACROSS THE OSSEOINTEGRATED INTERFACES OF DENTAL IMPLANT

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A modelling scheme for the stress analysis taking into account load transfer characteristics of the osseointegrated interfaces between dental implant and surrounding alveolar bone was investigated. Main aim was to develop a more realistic simulation methodology for the load transfer at the interfaces than the prefect bonding assumption at the interfaces which might end up the reduced level in the stress result.

In the present study, characteristics of osseointegrated bone/implant interfaces was modelled with material nonlinearity assumption. Bones at the interface were given different stiffness properties as functions of stresses. Six different models, i.e. tens0, tens20, tens40, tens60, tens80, and tens100 of which the tensile moduli of the bones forming the bone/implant interfaces were specified from 0, 20, 40, 60, 80, and 100 percents, respectively, of the compressive modulus were analysed. Comparisons between each model were made to study the effect of the tensile load carrying abilities, i.e. the effectivity of load transfer, of interfacial bones on the stress distribution. Results of the present study showed significant differences in the bone stresses across the interfaces. The peak stresses, however, were virtually the same regardless of the difference in the effectivity of load transfer, indicating the conventional linear modelling scheme which assumes perfect bonding at the bone/implant interface can be used without causing significant errors in the stress levels.

Key Words
Load transfer, Dental implant, Bone/implant interfaces

During months of initial adaptive period following an implantation operation, localized bone tissue grows to the metallic surface of dental implant at the threaded implant/bone interface. Apposition of osseous tissue leads to a close integration between the implant and the bone, i.e. osseointegration, which allows the bone to withstand the functional mechanical loads. As an implant or implant supported prosthesis begins to resist the masticatory force, however, stresses built up in the bone can cause resorption of the bone tissues especially around the neck part of the implant. According to earlier study by Adell et al, an average marginal bone loss of 1.5 mm was observed during the first year of the implant placement. The rate of annual bone loss was reduced to 0.1mm from the second year on.[1) In the recent studies, Olsson et al reported in 1995 that the mean marginal bone resorption was approximately 0.5 to 0.6mm after 3 years of function,[2) Lekholm et al, in 1999, 0.7mm,[3) and Naert et al reported in 2000 that the marginal bone loss during the first 6 months after abutment connection reached 0.71mm and then dropped to 0.036mm annually over a period of 10 years,[4) which means that the marginal bone loss decreases as the various systems are developed and surgical technique is improved.

One of the most important factors responsible for the loss of the implant supporting bone is stress.[5) Many factors, such as poor oral hygiene,[6) excessive strain,[7) and poor bone quality[8) have been suggested. Although concrete data are not available at the moment for a quantative evaluation, these factors, if allowed to exceed physiological level, can have catastrophic effects on the osseointegrated interface. When there is lack of osseointegration or a dental implant is not sufficiently anchored in the bone, even bigger mobility of implant would be the result since localized stress concentration can accelerate and aggravate the bone resorption process in a vicious circle.

A dental implant fixture act as an in-between element in the masticatory load transfer system, connecting the up above prosthesis and the alveolar bone. Osseointegration at the implant/bone interface plays an important role in the load transfer. Since excessive stress, by interfering the bone-remodelling mechanism, can destroy osseointegration, it is of particular importance to be able to predict and control the level of stresses in both the prosthesis and the bone. Therefore, various biomechanical studies have been attempted to evaluate the stresses in the dental implant related structures either by various experimental methods or by finite element analysis.[7,9-18)

Despite the importance of stress, however, the problem is not simple. When an implant supports a prosthesis with other implants or abutment teeth, the calculation of the load taken by a particular implant makes a big biomechanical challenge. Although when the load share of an implant can be decided rather straightforwardly, as is the case with a single implant restoration, an evaluation of the stresses needs a complicated modelling process. Many previous researchers, Moon and Yang,[7) Han, Jun et al,[13) Kim,[18) Jung,[19) Kim,[20) Meijer and Starmans et al,[21) and Skalak,[22) took rather qualitative approaches and have focused on removing intrabony stress concentrations through the adjustment of implant design shape, instead of calculating the intrabony stress quantitatively.

Although most of the above studies were carried out with various different implant systems; unequivocally most of them demonstrated the fact that cervix of the implants were the most vulnerable area in terms of stress concentration.[13,14,18) Meanwhile, Jung[19) came up with a numerical experiment scheme with the use of the finite element method to quantitatively predict the stress to initiate bone resorption at the cervix of an implant. Jung introduced a scheme of stepwise modelling of the bony pocket formation at the cervix of the implant. As the bony pocket deepened, stress concentration got relieved and peak stress area moved from the implant cervix to alveolar crest.

In the situation of direct contact between the bone and the implant, whose Young' s moduli are 5-10 times bigger than that of the former, a certain level of stress concentration is hardly avoidable. In the earlier stage, before physiologic adaptation stage, stress concentration should be more prominent. The finite element method is the only method to handle this complicated situation. Virtually no other method has been proven to be more useful in the evaluation of the stresses in these particular area. In order to analyse the stresses with acceptable precision in the stress results, however, F.E. models should be firmly based on physical realities. Unnecessary or oversimplified assumptions in the modelling procedure will only lead to a reduced level of precision and thereby insufficient reliability in the stress results. One key issue in the F.E. modelling of dental implants is the osseointegrated interface. The perfect bonding assumption which has so far been frequently employed has a risk of underestimation of the stresses. More realistic interface model is essential to further provide accurate stress analysis.

Lee,[23)] in an effort to improve the interface modelling, classified implant/bone interface into three characteristic areas and tried to simulate the load transfer characteristics of the interface. In Lee' s interface modelling, however, bone tissues need to be divided according to the dominant stress components. When an implant is subject to loads in multi-directions, which is the case when the implant is among the abutment members of a prosthesis, it would not be feasible to decide the dominant stress component at a single particular part of the interface.

In the present study, with the above in mind, a concept of tensile load transmittance factor at the interface was introduced. Different tensile and compressive stiffness were specified for the bones near the implant/bone interface to simulate the presumed differences in the stress transfer pattern between tensile and compressive loads. The characteristics of osseointegrated bone/implant interfaces was modelled with material nonlinearity assumption. The tensile moduli of the bones forming the bone/implant interfaces were divided into six different models, i.e. tens0, tens20, tens40, tens60, tens80, and tens100. Each model represented 0, 20, 40, 60, 80, and 100 percents, respectively, of the compressive modulus. Comparisons between each model were made to study the effect of the tensile load carrying abilities, i.e. the effectivity of load transfer, of interfacial bones on the stress distribution. Results of the present study showed rather significant differences in the bone stresses across the interfaces among different models.

MATERIALS AND METHODS

Many different types of implant systems have been developed and introduced during the last decade so that a suitable one for a particular patients can be rather easily found. One of the most reliable implant systems is Brånemark implant of which a standard type is 3.75 or 4.0mm in diameter. Wide type of 5 or 6mm, or a narrow type of 3.25mm in diameter can be selected as well depending on the quality of bone conditions. Naturally implants of wide diameters are advantageous over the narrow ones in that they have larger stress bearing areas thereby lowering the level of stresses, which is critical for dental implants to have longer span of life. In the present study, 3i standard diameter implant with 3.75mm diameter by 10mm in length, 3rd generation surface treatment, was selected as a study model.

1) Geometry of the Implant and the Bone

Higher stress concentration at the bone near the neck of an implant in the earlier stage after implantation can be gradually relieved as the bone adapts to the stresses and the soft tissues get to replace the bone. A lot of studies have aimed explicitly at clarifying the role of stresses on the amount of bone resorption. Jung,[19)] in an effort to quantify the stress criteria for the bone resorption, modelled 4 different stages of bone resorption, and observed that the levels of stress concentration decreases as a result of the bone resorption at the cervical area of the implant. As it happened, location of the stress peak moved to the alveolar crest from the neck area.

Most of the working implants therefore have a saucer shaped bone resorption near their cervical areas. In the present study, a typical one year old implant of about 1.7 mm deep cervical bony pocket as shown in Fig. 1. was chosen for analyses. A pocket of 1.7mm depth represents a state that the bone resorption has occured upto the second thread of implant fixture. It seems worthwhile to note that the surface treatment in 3i implants upto the 2nd thread is made for soft tissue, unlike those below the 2nd thread.

Thickness and density of the cortical and cancellous bones have a direct effect on the distribution of stress around the implant. Therefore it is important to understand exactly the characteristics of the bone when selecting the implant. On the bone, the specification of implant, i.e. material, size, surface condition, thread type, and etc ought to be dependent. Zarb classified the bone quality around implant into 4 different types, according to the thickness of the cortical bone. In the present study type II bone by Zarb' s classification of which the thickness of its cortical bone is 0.75mm, and clinically most prevalent, was selected for the bone modelling. Soft tissue was excluded from the whole analysis because of its lower load carrying capacity. Mechanical property data for the materials used in this study is shown Table I.

Fig. 1. Typical dental implant used as a single crown abutment. Saucer shaped bone resorption is observed at the cervical area.

2) Axisymmetric Finite Element Model

Finite Element Method(FEM) is practically the only methodology facilitate the evaluations of the stresses in such complicated geometries as dental implants. Various commercial programs have been developed and commercially available. In the present study, ABAQUS (version 5.8) running on a Hewlett Packard work station was used.

The same mesh model used by Lee[23] and Jung[19] for their axisymmetric linear analysis, was used in the present study in order to make the direct comparison of the results. ABAQUS input file was made using the existing node and element connectivity data of Lee and Jung' s axisymmetric mesh model, which were established on Display IV of NISA II. The element connectivity format of NISA II and ABAQUS for the same 8 node axisymmetric element are different from each other. Therefore, it was needed to converse NISA II data into ABAQUS data format. A FORTRAN program was written for the conversion. Finally, checking of the input mesh model as well as output display were carried out using ABAQUS CAE.

Table I. Mechanical properties (implant and bone)

Material	Young Modulus (GPa)	Poisson ratio	Strength (MPa)	Tensile yield stress (MPa)
Titanium	102.2	0.35	-	-
Cortical bone	13.7	0.3	72-76 (tensile) 140-170 (compressive)	60
Cancellous bone	1.37	0.3	22-28 (tensile)	-

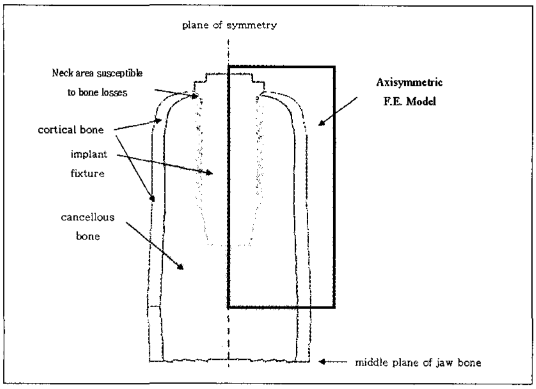

Fig. 2. Schematic diagram for a dental implant and an axisymmetric F.E. modelling. Soft tissues and/or the abutment, which are not included in the F.E. modelling are excluded in the diagram for clarity.

A code for nonlinear modelling was a must in order to model the nonlinear mechanical behaviour at the osseointegrated implant/bone interface. In particular, as was assumed in the present study, material nonlinearity of different tensile and compressive moduli could hardly be realized by other commercial codes than ABAQUS. In the present study, material nonlinearity was realized by using *Hyperfoam option in ABAQUS.

Axisymmetric model of the present study is shown in Fig. 2. In the processing 2-dimensional representation from the 3-dimensional structure, some simplifications in the modelling were required. The threads were simplified into the serrations and the spiral groove at the self tapping area was modelled with smooth surface. In the whole mesh, the CAX 8 solid elements of ABAQUS (8 node quadrilateral element) were used. Aspect ratio of all the elements was controlled not to exceed 5.0 in an effort to minimize the calculation error. The corner angles of elements were controlled to be within 45-135°.

As shown in Fig. 3, fine meshes were used at the implant /bone interface where the abrupt change of stress was expected. Meanwhile coarse meshes in the other areas were used to reduce computing time. The model consisted of 3967 elements and 10928 nodes in total. Only the bones surrounding the implants of which the height of 1.5 times as high as the implant length were included in the axisymmetrical mesh model. Some preliminary studies had revealed no significant stress pattern outside of this range.

In the whole analysis, the vertical load of 50N was applied at the top of the implant. As for the geometry boundary condition, as shown in Fig. 3, clamp condition was placed at the base plane.

3) Implant/Bone Interface Modelling

Enormous efforts have been made to strengthen the interface. In earlier stage implants were made without sophisticated surface treatment. In the next stage, or in the 2nd generation, coating materials such as hydroxyapatite was used for the purpose of accelerating the bone apposition and thereby osseointegration. The 3rd generation technology of blast and etched surface has moved quickly to the 4th generation surface. Here, the micropores are being introduced on implant surface in an effort to maximize the contact area and to result in highly complex three-dimensional mechanical interlocking.

When masticatory force is transmitted to the osseointegrated implant, the implant/bone interface plays an important role in the stress transmittance. Therefore, it is well worth taking into account the mechanical behavior here. For the increased precision in the stress analysis a realistic modelling scheme which takes into account the physical load transfer characteristics of the osseointegrated interfaces would be essential. In most of the biomechanical studies so far, perfect bonding has been assumed at the interface, which hardly reflects the osseointegration of Brånemark. Stresses might have been under estimated due to this kind of oversimplification.

By Brånemark, osseointegration was defined as a direct, functional and structural connection between implant and bone, without intervening soft tissue, clinical mobility nor any of peri-implant radiographical space. To reflect the characteristics of osseointegration in the F.E. Model, a contact mod-

Table II. Elastic moduli of the interfacial bones representing the effectivity in the load transfer across the implant/bone interfaces

Model	compressive modulus (GPa)		tensile modulus (GPa)	
	cortical bone	cancellous bone	cortical bone	cancellous bone
tens 0	13.7	1.37	0.13	0.013
tens 20	13.7	1.37	2.74	0.274
tens 40	13.7	1.37	5.48	0.548
tens 60	13.7	1.37	8.22	0.822
tens 80	13.7	1.37	10.96	1.096
tens 100	13.7	1.37	13.7	1.37

el looks appropriate at a first glance. In this case, however, the frictional mechanism, i.e. the frictional coefficient at the interface for example need to clarified by careful experimental measurement which ought to be an extremely complicate and difficult task.

In the present study, in order to simulate the mechanical behavior of the osseointegration, nonlinear stiffness properties at the interface were specified. The tensile moduli were specified as some fractions of the compressive modulus for the bones at the vicinity of the implants, to simulate the different stress transmit behaviour between tensile and compressive stresses at the interfaces. Using the fraction factors from 0 to 100% at the step of 20%, a total of six types of interface modelling, i.e. tens0, tens20, tens40, tens60, tens80, and tens100 were analysed to simulate the possible variation in the mechanical properties of the osseointegration at the interface.

Mechanical property data used for the interface models are shown in Table II.

RESULTS AND DISCUSSIONS

Vertical load, Fy, of 50N was applied to the center point of the axisymmetric model as shown in Fig. 3. As for the displacement boundary condition, constraints were placed for the displacement components in y direction, i.e. uy=0 across the bottom surface. The center point at the bottom surface was clamped to ensure stability of the boundary and of the analysis. It seems worth noting that in this axisymmetric model, x represents the radial direction and y the circumferential one. The same load and displacement boundary conditions are applied to all the six interface models. Vertical loads of 15-50N have frequently been suggested as normal loading range in proper masticatory functions.[9]

Loads of lateral or horizontal directions, and bending or twisting moments were shown to generate higher stresses than vertical loads. In clinical situation, however, those non-axial loading conditions should be avoided to ensure dental implants remain stable for length of time. Therefore it is appropriate to assume that vertical loads are the main loads for dental implants especially when used in single crown restorations.

Fig. 6(a), (b) are stress results for the case of tens0 model. It was assumed in tens0 that implant/bone interfaces are capable of transmitting only the compressive loads or stresses. Here, the mechanical interlocking at the interfaces shown in Fig. 4 were assumed to be incapable of transferring any of tensile stresses. This condition was simulated by specifying very small values of tensile modulus for the interfacial bones. In the present study, 0.1% of the compressive modulus is given for the tensile modulus. This means that the mechanical characteristics of the osseointegration at the interfaces are no more effective than a mere contact condition. Tensile modulus of small magnitude, not totally reducing to zero, is needed to ensure the stability of numerical calculation procedure. Null tensile modulus could cause divergence in the solving process as well as singularity problems.

The stress data shown by band plotting in figures from Fig. 6 to Fig. 11 are the principal stress III, i.e. the maximum compressive stresses. Equivalent stresses or von Mises stresses which have frequently been adopted for the evaluation of bone stresses in many previous studies,[13] however, are more to do with the shear stresses which might be useful when yielding of the materials is of concern. Stress which have been regarded as a main factor in dynamic bone metabolism, however, is the normal stress component. Two practical rules have long been used in orthodontic process, for example. Tensile stresses induce deposition in the alveolar bones whereas compressive ones are to do with resorption of the bones. To the knowledge of the author, there has been no hard evidence that von Mises stress can be adopted to decide bone metabolism.

Fig. 6(a) shows the stress distribution across the implant and the surrounding bones. It is observed that the stresses near the whole threaded part of the implant fixture are more or less the same magnitude of around 0.3-0.5 MPa. This suggests that rather a uniform or homogeneous load transmittance takes place at the whole threaded part.

It would be an extremely difficult task and that no previous study seemed to be able to clarify the

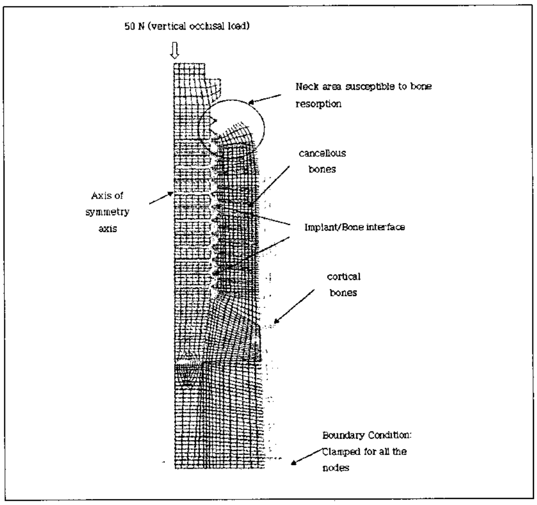

Fig. 3. Axisymmetric finite element modelling for the 3i dental implants subject to vertical occlusal load of 50 N and surrounding bones. Soft tissues and/or the abutment are not included in the F.E. modelling.

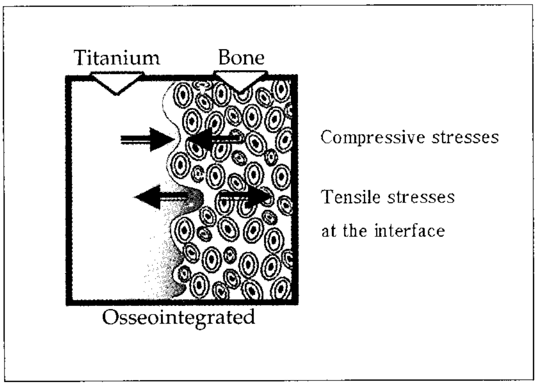

Fig. 4. Schematic diagram for osseointegrated implant/bone interfaces with mechanical interlocking . The effectivity in the load transfer varies depending on the kind of the loads or stresses.

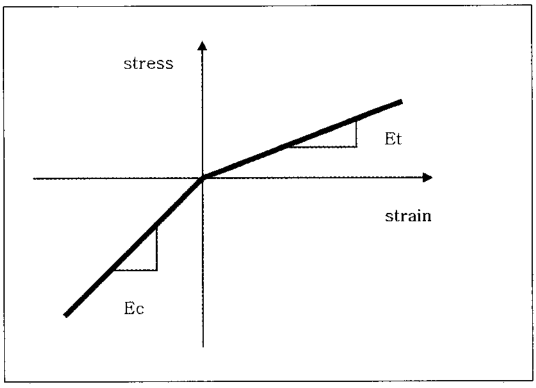

Fig. 5. Schematic diagram for material nonlinearity model (Ec: compressive modulus, Et: tensile modulus). Et / Ec = 0, 0.2, 0.4, 0.6, 0.8, 1.0

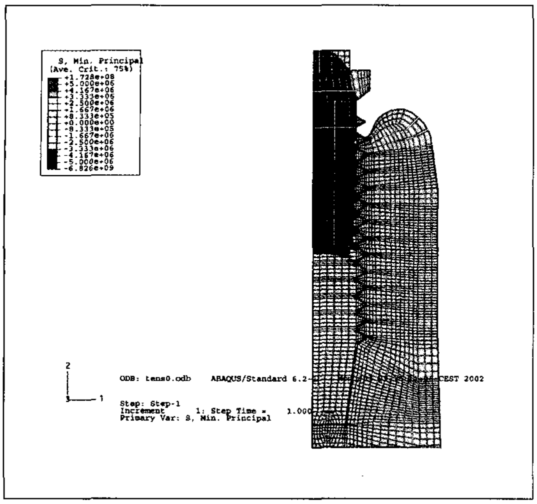

Fig. 6(a) Stress distribution across the implant and bone (model : tens 0)

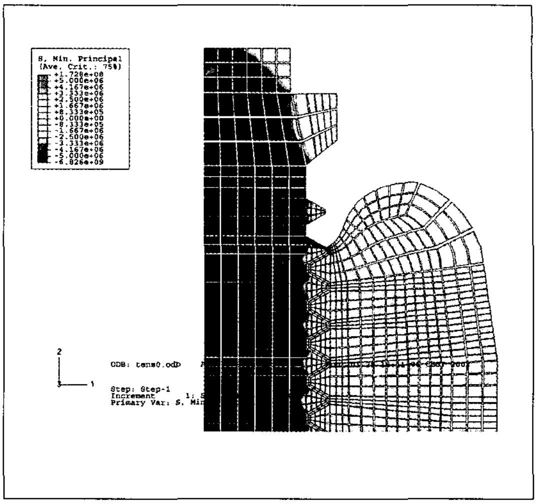

Fig. 6(b) Magnification of the neck area

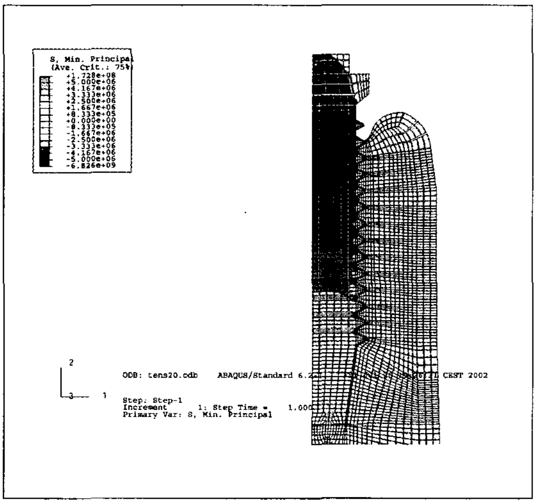

Fig. 7(a) Stress distribution across the implant and bone (model : tens 20).

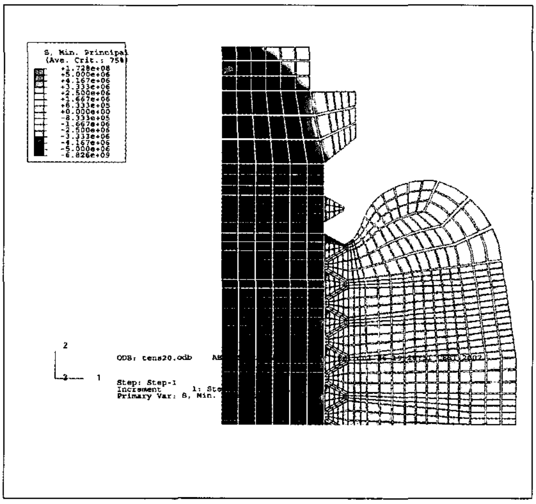

Fig. 7(b) Magnification of the neck area

stress condition which can possibly initiate bone resorption under normal masticatory situation. Jung,[19] by an incremental modelling of the process of the bone resorption taking place at the cervix of the implant suggested that the level of compressive stresses need to be of 2.5-3.5 MPa for the resorption process of alveolar bones to take place. Stresses of 0.3-0.5 MPa near the screw part therefore are thought to be well within safe level. The peak stress near the cervical area of the implant as shown in Fig. 6(b), however, is shown to be as high as 3.3 MPa.

Figures from Fig. 7(a), (b) to Fig. 10(a) and (b) are the results for tens20, tens40, tens60, and tens80, respectively. General remark about the stresses is that as the tensile stress transmittance ratio at the interface gets higher the level of the bone stress gets

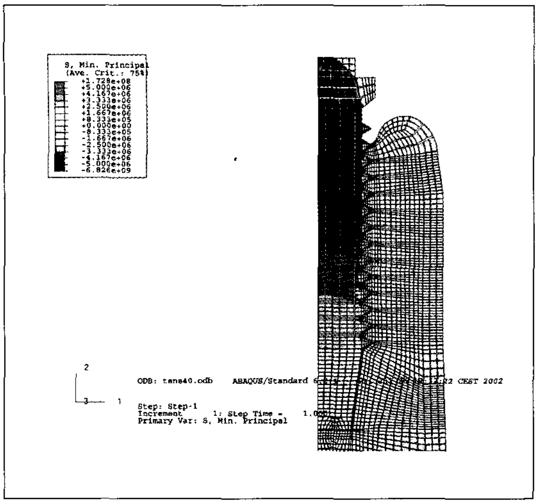

Fig. 8(a) Stress distribution across the implant and bone (model : tens 40)

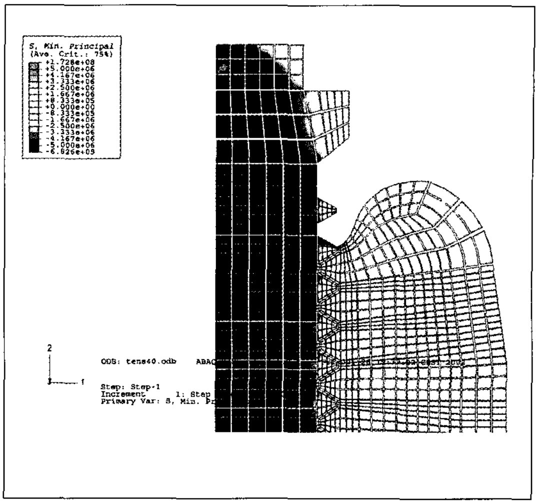

Fig. 8(b) Magnification of the neck area

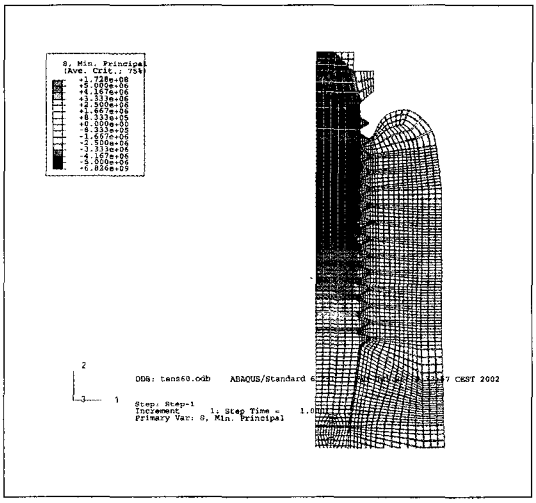

Fig. 9(a) Stress distribution across the implant and bone (model : tens 60)

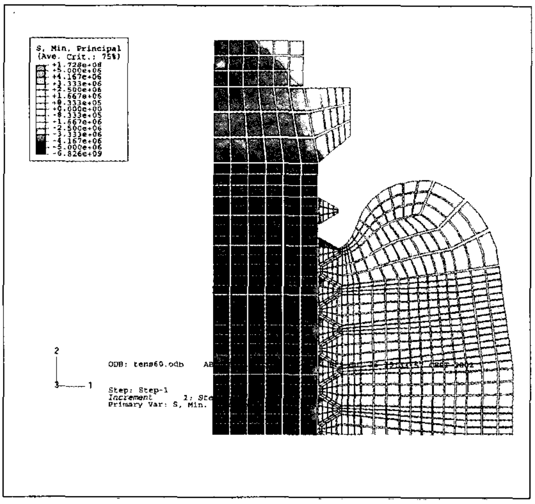

Fig. 9(b) Magnification of the neck area

smaller. The stress near the 7th screw in tens0 model, for example, is 0.37 MPa while stresses of approximately 0.25 MPa are found in tens20, tens40, tens60, and tens80 model. The higher tensile load transmittance ratio means the higher material effectivity and tend to produce lower stress.

In order to simulate the different load transfer characteristics at the implant/bone interface, Lee came up with a rather new idea for interface modelling. Lee analysed three different models to investigate effects of osseointegration at the interface between dental implants and surrounding bones on the stress distribution. Bone/implant interfaces on the screws of the implant were divided into three characteristic surfaces, i.e. compression dominant, tension dominant, and shear dominant ones. In the Bond

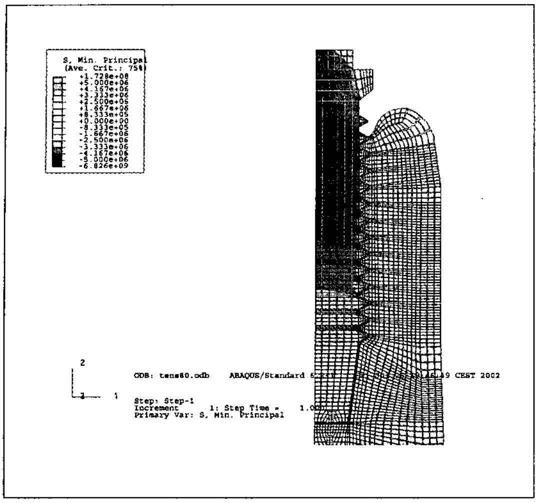

Fig. 10(a) Stress distribution across the implant and bone (model : tens 80)

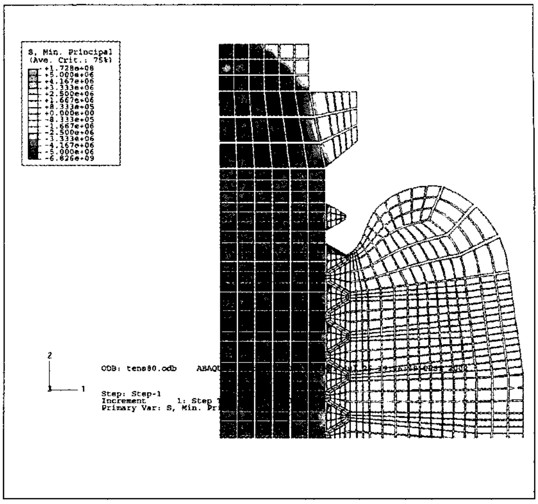

Fig. 10(b) Magnification of the neck area

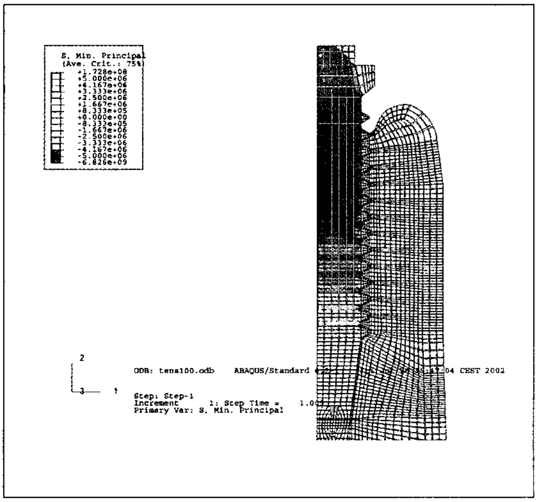

Fig. 11(a) Stress distribution across the implant/bone interface (model : tens100).

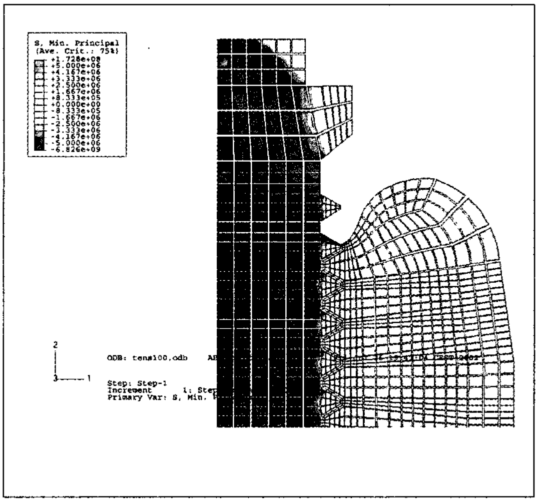

Fig. 11(b) Magnification of the neck area

model, a condition of perfect bonding was assumed at the whole interface. On the other hand, in the cases of S50T10 and S10T10, , the effectivities of shear and tensile load transfer were assumed to be different from that of compression load. In the present study, however, a nonlinear modelling scheme to avoid such a complicated process of dividing the implant/bone interface into several characteristic area which might be lacking of scientific evidence when the implants resist loads not just from vertical direction but from some other directions as well. When implants are subject to loads from various directions, stress near the implants will become extremely complicated so that it would not be straightforward to classify the interface into several characteristic areas where a particular stress component is domi-

Table III. Comparison of stresses at the bone/implant interfaces in each of the six models.

Location (serration no.)	tens0	tens20	tens40	tens60	tens80	tens100	ele. No.
1	3.91E+05	3.09E+05	3.01E+05	2.97E+05	2.95E+05	2.94E+05	1213
2	3.34E+05	2.84E+05	2.68E+05	2.62E+05	2.58E+05	2.56E+05	1201
3	3.40E+05	2.64E+05	2.47E+05	2.41E+05	2.38E+05	2.36E+05	1081
4	3.47E+05	2.56E+05	2.39E+05	2.32E+05	2.29E+05	2.27E+05	1093
5	3.55E+05	2.53E+05	2.35E+05	2.28E+05	2.25E+05	2.23E+05	1105
6	3.63E+05	2.53E+05	2.34E+05	2.27E+05	2.23E+05	2.21E+05	1117
7	3.72E+05	2.54E+05	2.34E+05	2.27E+05	2.23E+05	2.21E+05	1129
8	3.83E+05	2.58E+05	2.37E+05	2.29E+05	2.24E+05	2.22E+05	1141
9	3.97E+05	2.65E+05	2.43E+05	2.34E+05	2.30E+05	2.27E+05	1153
10	4.18E+05	3.04E+05	2.88E+05	2.80E+05	2.77E+05	2.74E+05	1165
11	4.40E+05	2.87E+05	2.62E+05	2.52E+05	2.47E+05	2.45E+05	1177
12	4.95E+05	3.13E+05	2.84E+05	2.73E+05	2.67E+05	2.64E+05	1189
13	5.70E+05	6.36E+05	6.20E+05	6.12E+05	6.08E+05	6.06E+05	1222
self tap	2.79E+05	4.82E+05	4.94E+05	5.00E+05	5.03E+05	5.04E+05	3810
apex	8.25E+05	8.19E+05	7.96E+05	7.88E+05	7.84E+05	7.82E+05	3841
Min.(neck)	1.18E+07	5.11E+06	4.85E+06	4.77E+06	4.73E+06	4.71E+06	3398
Min.(vert)	2.26E+06	1.76E+06	1.75E+06	1.74E+06	1.74E+06	1.75E+06	3911

nant over others. It would be virtually impossible to handle this type of situation manually. Therefore it is understood that Lee' s scheme might work in the case when axial loads are the only load of concern. The advantage of a nonlinear modelling scheme is that it can be applied to implant/bone system regardless of loading conditions. Using nonlinear modelling, mechanical characteristics of the bone materials near the interface are rather intrinsically but automatically decided according to the stress level.

Fig. 11.(a) and (b) are the result for tens100 model in which true bonding, i.e. adhesion, was assume at the implant/bone interface. The peak stress at the alveoler crest of approximately 2.5 MPa is not different from ones in the tens0 and other models. Nevertheless, the stresses near the body part of the implant are 25% lower than those in the case of tens0, indicating that a significant under evaluation of stress as a result of over simplification of the mechanical behaviour of the interface.

CONCLUSION

It is well recognized that the finite element method is the only tangible option to evaluate the stresses in the bones surrounding dental implants. Virtually no experimental techniques has ever been successfully employed to measure the stresses in these particular area. In order to analyse the stresses with acceptable precision in the stress results, however, F.E. models need to be firmly based on physical realities. Unnecessary or oversimplified assumptions in the modelling procedure will only lead to a reduced level of precision and reliability in the stresses results. Despite the numerous researches and refinements in this area, one key issue in the F.E. modelling of dental implants still remains: the interface problem. The perfect bonding assumption has so far been frequently employed at the risk of underestimation of the stresses. To date, there has been very little improvement in the interface modelling scheme. Use of more reliable interface model is needed to improve the precision in stress analysis.

A new modelling scheme for the load transmit behaviour taking place at the dental implant/bone interfaces, was investigated. Nonlinear stiffness properties, i.e. the tensile modulus was specified as some fractions of the compressive modulus for the bones at the vicinity of the implants, to simulate the different stress transfer behaviour between tensile and compressive stresses at the interfaces. The fraction factors were given from 0 to 100% at the step of 20%. Total six types of interface modelling, i.e. tens0, tens20, tens40, tens60, tens80, and tens100 were analysed to simulate the possible variation in the mechanical properties of the osseointegration at the interface. The numbers following the word 'tens' stands for the tensile modulus in percent of compressive modulus, which in turn represent the effectivities of tensile stress transfer. Comparisons between each model were made to study the effectivity of tensile load transfer of interfacial bones on the stress distribution. Within the scope of the present study, the following conclusions were drawn.

- In axisymmetrically modelled bones surrounding the standard 3i implants of 3.75 mm diameter, with initial bone loss of 1.7mm at the cervix of the implant, significant differences were observed in the bone stresses across the interfaces between the six different interface models. Some 50-80% differences in the stresses level at the interface bones near the screws between tens0 and tens100 models were observed. This means that underestimation with as much order are likely in the stresses results calculated from perfect bonding interface models.
- The differences in the peak stresses which play more important role clinically, however, were found to be not as much influenced by the difference in the tensile moduli. The differences in the peak stress which were observed near the alveolar crest was less than 10%. Therefore, it might be concluded that unacceptable errors are *not likely using the perfect bonding interface* model between dental implants and the surrounding bones when a sufficient bone loss had taken place at the neck area.

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