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3–D Numerical Analysis of the Stress State Caused by Short-Term Loading of a Fixed Dental Implant containing a "PDL-Like" Nonlinear Elastic Internal Layer

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Abstract: We study the mechanical behavior of a prototype osseointegrated dental implant containing a thin internal layer, designed in such a way as to simulate the existence of the periodontal ligament (PDL). Experimental stress-strain curves suggest that the behavior of the PDL can be simulated by means of a compressible hyperelastic constitutive model, at least for short-term loading. We have adopted one such a model to describe the mechanical behavior of the internal layer in the prototype implant design, studied by means of several 3–D Finite Element analyses. The results indicate that the presence of such a nonlinear internal layer is quite significant, in terms of stress redistribution, specially for all the loading/boundary conditions involving a strong static indeterminacy. It remains still difficult to assess whether the stress redistribution produced by the studied implant is beneficial in terms of bone behavior, owing to the lack of knowledge of the real mechanical fields which develop in the tooth-PDL-bone system under loading.

keywords: Osseointegrated dental implants; periodontal ligament; nonlinear finite element analysis.

1 Introduction

This work is motivated by the observation that, in a purely mechanical sense, the commonly adopted designs of fixed, osseointegrated dental implants appear to be quite poor. By this we do not intend to challenge the clinical success of such designs (difficult to quantify with precision, however), but to simply state that they are far from "optimal", owing (i) to their very high stiffness and (ii) to the search for full osseointegration.

From a strict mechanical viewpoint we can define as optimal an implant which, after obviously fulfilling the basic mechanical and biological standards for its own self, causes in the surrounding bone, upon loading, stress and strain states equal to those existing in the natural arrangement (tooth plus periodontal ligament — hereafter shortened as PDL — plus bone). Taking this viewpoint, it is obvious that practically all the conventional implant designs used in the professional practice are not optimal.

A list of "non-optimal" features of conventional implants includes the following items:

1. the strong contrast between the stiffnesses of bone and implant, responsible of a non-natural stress distribution in the bone;
2. the absence of the periodontal ligament, responsible of both a non-natural stress distribution and of an utterly non-natural prosthesis mobility;
3. the existence of threads, frequent in practice, both at the bone-implant interface and internally (connection screws). Threads act inevitably as a source of stress concentration, non-natural in the surrounding bone as well as very dangerous in terms of life of the implant itself, a mechanical part subjected to repeated cyclic loading (see for instance Genna, 2003);
4. the possible presence of self-stresses due to geometric misfits (Pietrabissa et al., 2000; Rangert and Renouard, 1999);
5. the remarkably high working stress level, quite close to the yield limit of the commonly adopted alloys (gold or titanium), usually accepted for the internal connection screws, arising essentially as a consequence of the screw tightening.

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The primary responsible for the large difference between the behavior of a healthy tooth-PDL-bone system and one including an osseointegrated implant is the absence, in the latter, of the PDL. Indeed, the concept of introducing a "soft" component into the design of a dental implant, to somehow compensate for the absence of the PDL, is not new. Rieger et al. (1989) propose the use of a "soft" implant altogether, in order to minimize the stresses in the bone. Buser et al. (1990) have explored experimentally the possibility of inserting an implant directly into a retained apical root tissue, in such a way as to maintain the PDL around the root and possibly develop new cementum and new connective tissue fibers around the implant. The design of an implant including a stress-absorbing, non-void element inside the fixture structure has been studied, at a very simple numerical level, by van Rossen et al. (1990), but has apparently received little further attention. The use of a so-called "intra-mobile element", made of soft material, placed between fixture and abutment, is the recipe of the IMZ implants, studied by several authors (Richter et al., 1990; Holmes et al., 1992; Lill et al., 1988, among others) and used in practice. Mejier et al. (1995) analyze a design including a soft layer placed at the interface between bone and implant. Clift et al. (1995) study the insertion of a "flexible internal post" in the fixture, an empty space designed in such a way as to cause a stress redistribution in the jaw bone, upon loading (specially transversal), from the neck of the fixture towards the inner parts of the bone.

All this effort does not seem to have caused much practical effects. We feel there are several reasons, beside those connected with clinical, biological, and technological factors, for looking with caution to the results presented in the above-quoted work. No "in vitro" experiment can support a new implant design, which, therefore, before a clinical application can only be evaluated by means of numerical simulations. In most of the work summarized above, the numerical (Finite Element) models adopted are two-dimensional, based on the assumption of linear elasticity for all components; they are therefore inadequate both in terms of geometry (see, for instance, Corradi and Genna, 2003) and of material modeling. The real behavior of the PDL, which the "soft" components should obviously reproduce to bring the implant closer to optimality, is in fact nonlinear viscoelastic. In particular, the nonlinearity of the stress-strain curve of the PDL, even for short-term loading, is very strong, as shown both by the mobility curves reported for instance in Parfitt (1960), and by the experimental work done by Ralph (1982), Pini (1999) and Pini et al. (2000).

Only the IMZ idea has produced widespread clinical applications. Nevertheless, the particular geometry of such a design makes it appealing more for stress *damping* than for stress redistribution purposes. The promising results obtained in terms of stress distribution in Richter et al. (1990) are based on a completely wrong numerical model but, most importantly, are obtained by defining the external actions on the implant as a prescribed displacement, which clearly overemphasizes the effect of the IMZ design, as noted also in Brunski (1992)[4]. If one applies a *force* on a IMZ implant, such as done, for instance, in Holmes et al. (1992), one finds very little difference in the stress state computed around the implant, specially in the case of purely axial loading, with respect to that computed for a standard design. The only obvious advantage of the IMZ implant is the improved mobility, which makes it appealing as a support for implants partially supported by natural teeth.

A general problem is that for all these analyses the touchstone is missing, i.e., the desired optimal (i.e., very similar to the real one around a healthy tooth) stress distribution in the bone. Therefore, even having the possibility of performing very refined numerical analyses, it still remains very unclear how the "optimum" implant design should transmit stresses and strains to the jaw bone around the implant, and any attempt at designing a modified implant, with the purpose of producing an "a priori" defined stress state, should be considered with great caution. The only feasible alternative appears therefore to try to construct an implant resembling as closely as possible the "natural" configuration, i.e., the tooth-PDL system.

To this purpose the basic idea proposed in van Rossen et al. (1990) seems the most promising. In this work we adopt precisely such an idea and, starting from a standard implant design (threaded fixture, connection screw and abutment), we propose a prototype modified design, whose novelty consists in the inclusion within the fixture of a soft layer made by a material resembling as

[4] It is worth noting that it is not yet clear whether the action on a single tooth/prosthesis, deriving from the masticatory force exerted by the jaw muscles, is more a force or a displacement (this depends on the stiffness of all the parts involved in such a process; it would be quite an important task, albeit difficult, to try and quantify this aspect).

closely as possible the PDL. We study numerically, by means of 3–D, nonlinear Finite Element analyses, the effect of several load types on the stress state both in the implant and in the surrounding bone. Particular attention is given to the choice and description of the material to be used for the internal soft layer acting as a stress-absorbing/redistributing device.

The choice of such a material should be guided by any available data concerning the mechanical behavior of the PDL. The lack of such data has so far led to try and study the PDL, in numerical models, as a *linear elastic* material with suitably chosen elastic moduli. At the light of the strong nonlinearity of the PDL behavior, such an idea appears difficult to apply, as proved by the information, given in Rees and Jacobsen (1997), that literature values for Young's modulus of the PDL range from 0.07 to 1750 MPa — a clear indication of inadequacy of the linear elasticity assumption. In the same paper the indication can also be found that the PDL should be treated as an incompressible or almost incompressible material, with a suggested Poisson's ratio usually greater than 0.45. Only few authors (Andersen et al., 1991; Williams and Edmundson, 1984) leave space for different assumptions.

Very few experimental results are available concerning the mechanical behavior of human PDL. One of these is given in Ralph (1982), but only in terms of force-displacement curves, with a tensile strength found at an average of 2.4 MPa. More information is given in recent work by Pini (1999) and Pini et al. (2000), which describes the results of experiments performed on *bovine* PDL. Uniaxial stress-strain curves are reported, both for tension/compression and shear tests. These results confirm the qualitative observations which can be deduced from the earlier experiments of Parfitt (1960) and Ralph (1982), and indicate that even for short-term loading the PDL behaves as a highly nonlinear medium. Moreover, there is a strong suggestion that the PDL should be considered as a compressible material, even if the same author furnishes, in this respect, contradictory information (Pini, 1999; Natali et al., 2000).

Here we study the effect of an internal layer of a nonlinear "PDL-like" material, defining its properties in such a way as to match typical experimental stress-strain curves among those reported in Pini (1999) and Pini et al. (2000). This work is only a first approach to the study of a modified implant containing a *nonlinear* stress-absorbing element, and it is aimed only at highlighting the purely mechanical aspects implied by the presence of such a device. Several other extremely important topics will not be considered here, most notably (i) biological/biocompatibility aspects; (ii) viscous effects or, in general, long-term loading effects; (iii) dynamic effects, whose analysis would require, as an essential factor, the damping properties of the PDL-like layer, very difficult to assess; (iv) technological problems.

It is also important to make it clear that in this work we do not suggest a specific material to be used for the stress-absorbing internal layer (some possible ideas will be given, however, in the sequel), but we will *assume* to have available a material whose mechanical, short-term behavior is essentially analogous to that experimentally measured for bovine PDL.

Only a single, freestanding implant will be studied, even if, as obvious, and pointed out by several authors, one of the most beneficial effects of a correctly designed intramobile element of any shape would be noted in the case of prostheses partially supported by natural teeth. This case requires, in our opinion, a careful analysis in itself, since, in the case of multiply supported prostheses, a single supporting fixture could be subjected to loads which can hardly appear on a freestanding implant, such as purely tensile axial ones. This issue is outside the scope of the present work and is the subject, together with others, among those quoted above, of work in progress.

2 Geometry, Materials, and Loading of the Modified Implant

Figure 1 shows a section of the Finite Element model of the studied modified implant. The modification, with respect to a standard implant design, is the thin layer (shown in dark blue in Figure 1), whose minimum thickness is of about 0.2 mm, made of a material matching as closely as possible the behavior of the PDL and, at the same time, having enough strength to survive the applied external actions for several millions of cycles.

As starting data, for the choice of such a material, we have used the experimental curves, obtained for bovine PDL, given in Pini (1999) and Pini et al. (2000). Two such curves, giving the uniaxial tension/compression and shear stress-strain behavior, respectively, are shown in Figure 2. The mathematical modeling of the behavior described by the curves of Figure 2 is a very difficult task, specially in a continuum, three-dimensional context. In

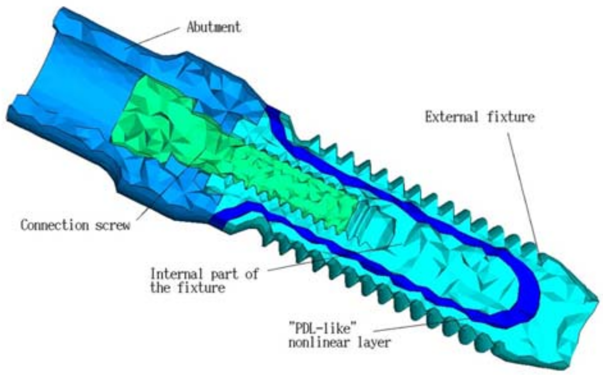

Figure 1 : Sectioned view of the modified implant design. In the standard design the internal layer (in dark blue) is absent.

fact, Gei et al. (2002) have proposed to simulate the presence of the PDL in healthy teeth by means of *interface* finite elements. In this paper we are interested in details of the stress state also inside the PDL-like layer, and, as a consequence, we must model it by means of continuum finite elements, requiring a continuum constitutive model. Previous work based on the same experimental results proposes either the use of an ad-hoc defined hyperelasticity law (Pini, 1999; Pietrzak, 1997; Pietrzak et al., 1998) or, more simply, the use of the Ogden hyperelastic, incompressible constitutive model (Natali et al., 2000).

In this work we have used a hyperelastic constitutive law allowing for compressibility of the material. This has been done both on the basis of the conclusions reached in Pini (1999), and to avoid the inevitable numerical problems arising when straining too much a thin layer of incompressible material. We have used the compressible hyperelasticity model of Storåkers (1986), based on the following equations.

Denoting with **F** the deformation gradient (we use here a standard continuum mechanics symbology, explained in detail, for instance, in Malvern, 1969), with J its determinant, with **B** the symmetric left Cauchy–Green deformation tensor

$$\mathbf{B} = \mathbf{F}\mathbf{F}^t \tag{1}$$

(the symbol ^{t} denotes the matrix transposition operation), with I_1 and I_2 the first two invariants of tensor **B**, defined

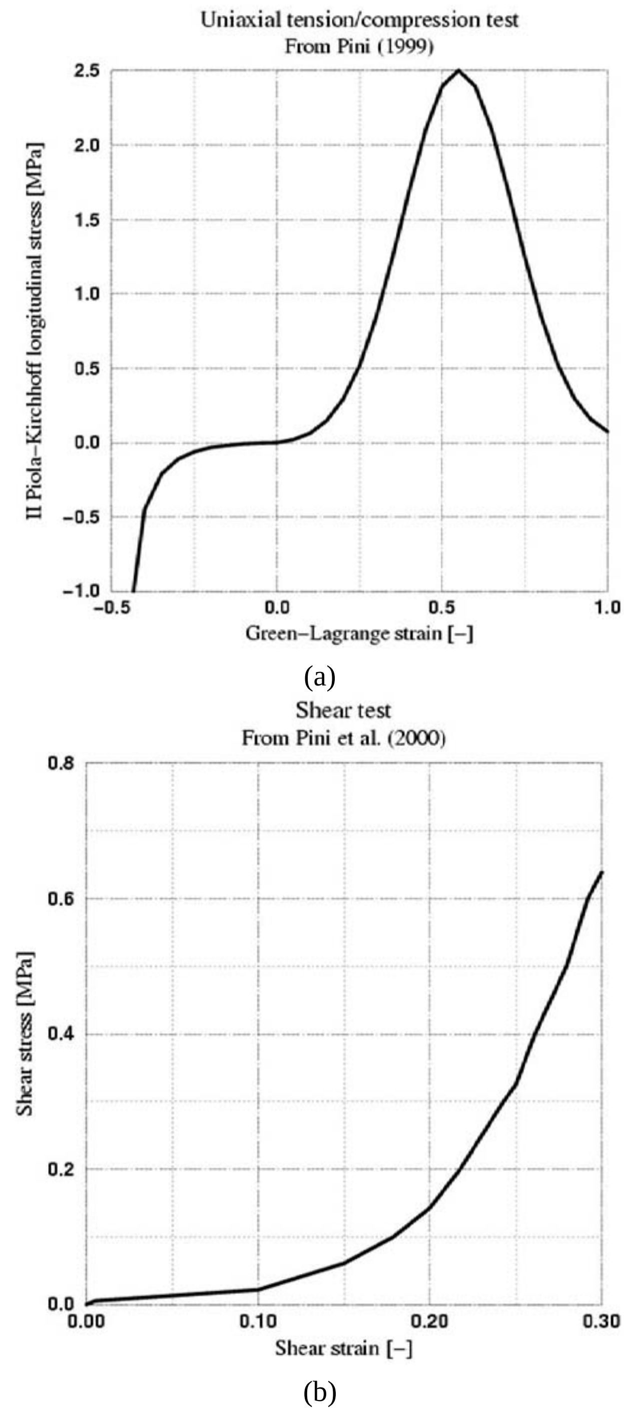

Figure 2 : Experimental uniaxial stress-strain behavior of bovine periodontal ligament (from Pini, 1999; Pini et al., 2000): (a) tension/compression; (b) pure shear.

as

$$I_1 = tr[\mathbf{B}];\; I_2 = \frac{1}{2}\left(I_1^2 - tr[\mathbf{BB}]\right) \tag{2}$$

(the symbol $tr[\cdot]$ indicates the trace of a tensor, i.e., the sum of its diagonal terms), this model assumes the existence of a strain energy potential $\mathcal{U}$ of the form

$$\mathcal{U} = \mathcal{U}(I_1, I_2, J) \tag{3}$$

from which the Cauchy stress τ is obtained as follows:

$$\tau = \frac{2}{J}\,\mathrm{DEV}\left[\left(\frac{\partial\mathcal{U}}{\partial I_1} + I_1\frac{\partial\mathcal{U}}{\partial I_2}\right)\mathbf{B} - \frac{\partial\mathcal{U}}{\partial I_2}\mathbf{BB}\right] + \frac{\partial\mathcal{U}}{\partial J}\mathbf{I} \tag{4}$$

where the symbol DEV$[\cdot]$ indicates the deviatoric part of its argument, and $\mathbf{I}$ is the unit tensor. The Storåkers model assumes a strain energy function $\mathcal{U}$ defined in terms of principal stretches $\lambda_{I,II,III}$ as follows (neglecting thermal effects):

$$\mathcal{U} = \sum_{i=1}^{N}\frac{2\mu_i}{\alpha_i^2}\left[\lambda_I^{\alpha_i} + \lambda_{II}^{\alpha_i} + \lambda_{III}^{\alpha_i} - 3 + \frac{1}{\beta}\left(J^{-\alpha_i\beta} - 1\right)\right] \tag{5}$$

where α_i and μ_i are N parameters, depending on the material choice, which must be determined from experiments. The further material parameter β is related to Poisson's ratio ν by the following relation:

$$\nu = \frac{\beta}{1+2\beta} \tag{6}$$

in which $\nu \to 0.5$ implies $\beta \to \infty$.

There is no need here to go into more details of this model, available in the library of the Finite Element code ABAQUS (Hibbitt et al., 2001). The calculation of its parameters has been done by prescribing a best fit with the experimental uniaxial curves of Figure 2; to obtain a solution of this best fit problem we had to stop to $N = 3$ terms in the sum of eq. (5), and the corresponding parameter values are the following:

$$\alpha_1 = 7.904; \qquad \alpha_2 = 15.28; \qquad \alpha_3 = -5.7$$
$$\mu_1 = -0.08838; \qquad \mu_2 = 0.2394; \qquad \mu_3 = -0.05693$$
$$\nu = 0.35$$

Note that, in order to match the experimental curves of Figure 2, the material, governed by the Storåkers model and assumed to be isotropic, is definitely compressible.

With these numerical values of the parameters the Storåkers hyperelastic model becomes unstable (and therefore amenable to yield multiplicity of solutions or no solution at all) for deviatoric nominal strains of the order of 0.1 and volumetric nominal strains of the order of unity: this is a warning about its practical applicability over a wide range of loads.

The other parts of the implant are made by either pure grade 3 titanium or titanium alloy (Ti-6Al-4V), characterized by the following material parameters, obtained in our laboratory from uniaxial tension tests:

- grade 3 titanium: Young's modulus $E = 106000$ MPa; Poisson's modulus $\nu = 0.31$; yield stress $\sigma_y = 242$ MPa; tensile strength $\sigma_0 = 1045$ MPa; uniaxial plastic strain at failure $\varepsilon_u^p = 0.025$;
- titanium alloy Ti-6Al-4V: Young's modulus $E = 114000$ MPa; Poisson's modulus $\nu = 0.31$; yield stress $\sigma_y = 852$ MPa; tensile strength $\sigma_0 = 939$ MPa; uniaxial plastic strain at failure $\varepsilon_u^p = 0.045$.

We modeled these materials as elastic-plastic, governed by a hardening von Mises yield criterion. In defining the internal friction between the various contact surfaces we have used a friction coefficient $\mu = 0.40$ as suggested in Sakaguchi and Borgersen (1995). The interface between bone and implant has been considered fully osseointegrated, i.e., the two surfaces have been modeled as fully connected to each other.

Figure 3 shows the complete numerical model of the jaw-implant system. This model, despite its complexity (due to its own geometry, the nonlinearity of materials, the existence of unilateral contact with friction, different length scales, etc.), is still quite crude. Its main approximations consist in the boundary conditions (the lateral edges of the bone are fully fixed) and in the lack of simulation of either the teeth adjacent to the implant or their alveoli. Corradi and Genna (2003) have discussed the significance of these and other assumptions; since we are here interested essentially in the comparison of the stress state around the implant arising from two implant designs, we have not pursued the goal of defining a really "good" model. It is important, however, to recall that the lack of modeling of adjacent teeth/alveoli undoubtedly reduces the validity, in an absolute sense, of the results presented in the following Section, even if they should definitely be significant in terms of comparison between the two designs examined.

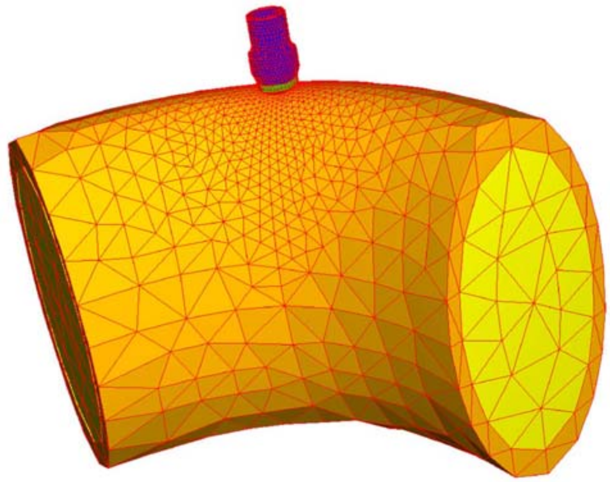

Figure 3 : External view of the full Finite Element mesh of the implant inserted in a model of a portion of the jaw.

We have used the following material data for the bone, considered as isotropic linear and elastic (from Meijer et al., 1995):

- cortical bone: Young's modulus $E = 13700$ MPa; Poisson's coefficient $\nu = 0.3$;
- spongious bone: Young's modulus $E = 1370$ MPa; Poisson's coefficient $\nu = 0.3$.

The model of Figure 3 has been discretized into 99561 4−noded tetrahedra, for a total number of 23909 nodes. The total number of unknowns, including Lagrange's multipliers arising from the unilateral contact description, amounts to 70560. Such a model has been studied under three different loading conditions, all defined in terms of *forces* applied on the top of the abutment:

1. a purely axial load of 300 N;
2. a purely transversal load of 20 N;
3. a combined load whose axial component is equal to 300 N and transversal component is 150 N. The transversal component, orthogonal to the axis of the implant, is included in the sagittal plane, directed from the lingual to the labial side.

These load values are average values among the huge variety of data available in the literature. It is safe to say that (i) the axial load value is quite high for a single implant, and is used here essentially to highlight the behavior of the studied system under limit conditions; (ii) likewise, the accompanying transversal action of case 3, equal to one half of the axial one, thus corresponding to a resultant force inclined by 30° from the axis of the implant, is an extreme case. In particular, it appears to be almost impossible to have such high forces — specially this transversal component — on a single freestanding implant replacing an incisor; again, these actions have been chosen as limit conditions; (iii) the intensity of the purely transversal load defined by condition 2 above, on the contrary, should represent, according to several authors (for instance, Brunski, 1992), a reasonable value for such an action. Note also that the use of a force type loading condition is expected to produce differences, between the standard and the modified implant designs, smaller than those caused by a mixed or a pure displacement one; thus, the results presented in the next Section must be considered as "conservative".

All the above-listed loading conditions are applied after a first loading step in which the tightening of the internal screw is prescribed. This preloading is here simulated by applying, inside the screw, a self-equilibrated, axial tension stress of about 500 MPa, corresponding, for the studied screw type (M2), to a tightening torque of 30 Ncm.

All these loads are defined as short-term static loading. No dynamic effects are taken into account, even if their analysis, in the presence of a stress-absorbing layer, would be quite interesting. No viscous effects, essential, for instance, in the analysis of orthodontic loads, are considered either.

The Finite Element analysis has been performed by means of the commercial code ABAQUS (Hibbitt et al., 2001), taking into account unilateral contact with friction and in a regime of large displacements and large strains. This last option, clearly useless for all the stiff parts of the system, is forced by the use of the hyperelastic constitutive law describing the behavior of the internal layer. All the nonlinearities (plasticity, hyperelasticity, contact, large strains) are dealt with by means of a Newton–Raphson iterative solution scheme, with no particular care taken for handling possible instabilities which, however, have had no apparent effect on the analyses reported in the sequel.

3 Finite Element Analysis Results

The results described in this Section refer to the comparison of the mechanical fields arising both in the implant and in the bone, under loading, between the basic and the modified implant designs. The results are presented as contours of von Mises equivalent stresses (a scalar measure of the “global” stress intensity at a point), all shown in MegaPascals.

Figures 4 concern the implant only, and illustrate the results of the analyses in an axial section of the implant. Figures 4a and 4b refer only to the effect of the screw preloading due to its tightening. Figure 4a shows results for the standard design, and Figure 4b the corresponding ones for the modified design. The effect of the soft internal layer is of substantially relieving the stresses in the screw, at the price of somewhat increasing the stresses in the internal part of the fixture. This part, however, is not much more stressed than in the standard design; note that there is also the possibility, if deemed necessary, of using a material with a higher strength than the grade 3 titanium considered here, since the internal part of the fixture does not contact the bone, and, therefore, it has no requisites of good osseointegration (even if, from the corrosionistic viewpoint, bimetallism might be a disadvantage). In any event, the peak stress of about 500 MPa is within the working range of the material. The external part of the fixture, connected to the bone, is practically unloaded, which is not the case with the standard design.

Figures 4c (standard design) and 4d (modified design) refer to the axial loading condition. The situation for the screw is practically unchanged with respect to the previous loading condition; it is apparent that, despite a moderate increase of the stress in the internal part of the fixture (from 400 to 500 MPa), the modified fixture design is able to withstand this load. The same comments apply to the results for the remaining loading cases (purely transversal and mixed), not shown for the sake of brevity.

The addition of the soft layer within the fixture causes a reduction by a factor of 2 of the magnitude of the plastic strains (from 0.005 to 0.0025) in the first two loading conditions, and an increase by the same amount (from 0.025 to 0.045) for the third one. These plastic strains are localized close to the threads of the connection screw, and are to be looked with some attention even in the standard design, since they might be the cause of low-cycle fatigue in these parts (Genna, 2003).

The subsequent Figures (5, 6, and 7) show the stress state in the bone surrounding the implant, always in terms of a comparison of von Mises stress contours between the two designs. These Figures do not show the effect of the tightening of the screw alone, but, in this respect, it suffices to say that the inclusion of the PDL-like layer in the implant design has the effect of practically annihilating all the self-equilibrated stresses in the bone, due to such an action; in the standard design, this same action creates *permanent* stresses whose peak value is of about 100 MPa.

Figures 5 refer to a top external view of the model; Figures 6 to a section of the model with a sagittal plane and Figures 7 to a section of the model with a frontal plane containing the implant axis, and orthogonal to the previous one. All the Figures refer only to the transversal and the mixed loading conditions, i.e., those creating the largest differences between the two designs.

In the case of the axial loading condition, in fact, the stress state in the bone is scarcely altered by the presence of the internal layer; the peak stress is of about 130 MPa in both designs, in the top surface of the cortical bone, caused by the threads. This substantial insensitivity of the peak stresses in the jaw bone to the presence of a stress-absorbing/redistributing device, *under purely axial loading conditions*, has been noted by several authors (for instance, Clift et al., 1995; van Rossen et al., 1990, etc.).

With reference to the results displayed in Figures 5 to 7 the following points are worth a comment:

- in the case of the purely transversal load the effect of the internal layer is quite significant. The maximum stress in the bone is reduced, by the presence of the internal layer, from about 90 to about 30 MPa, and all the other stress peaks associated to the standard design are likewise greatly reduced. For instance, the maximum stresses in the spongious bone are cut from 6.5 to 1 MPa. There are probably two reasons for this large difference, also with respect to the first loading condition: one is the increased static indeterminacy of the considered geometry, under the transversal loading condition, with respect to the purely axial one; a second is the wholly different stiffness exhibited by the PDL-like layer material under the low transversal load, with respect to that mobilized by the high axial load;

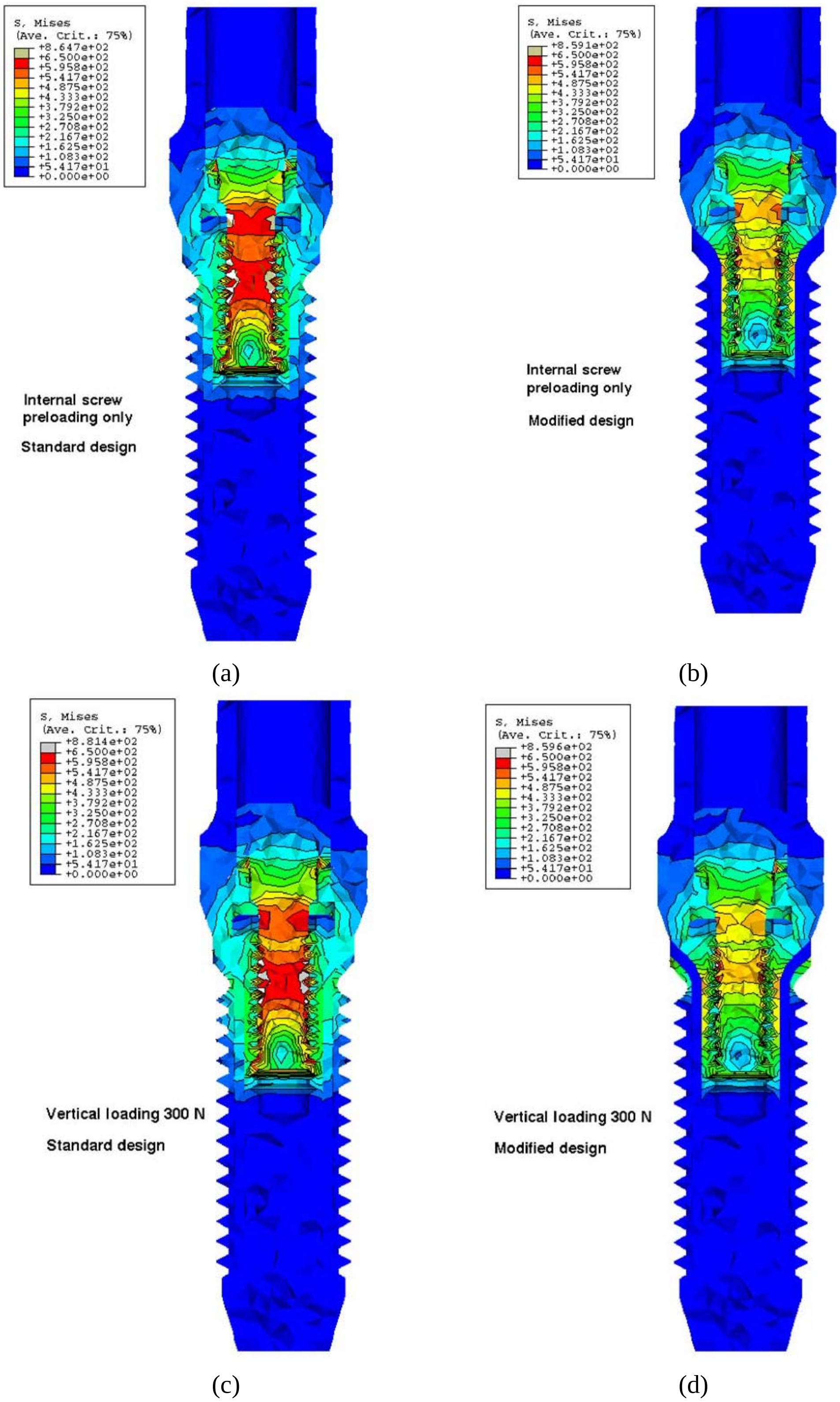

Figure 4 : Comparison of von Mises stress contours in an axial section of the implant for the standard and modified implant designs. (a, c) refer to the standard design; (b, d) to the modified design. (a, b) are for the internal screw preloading case only; (c, d) for the axial loading condition.

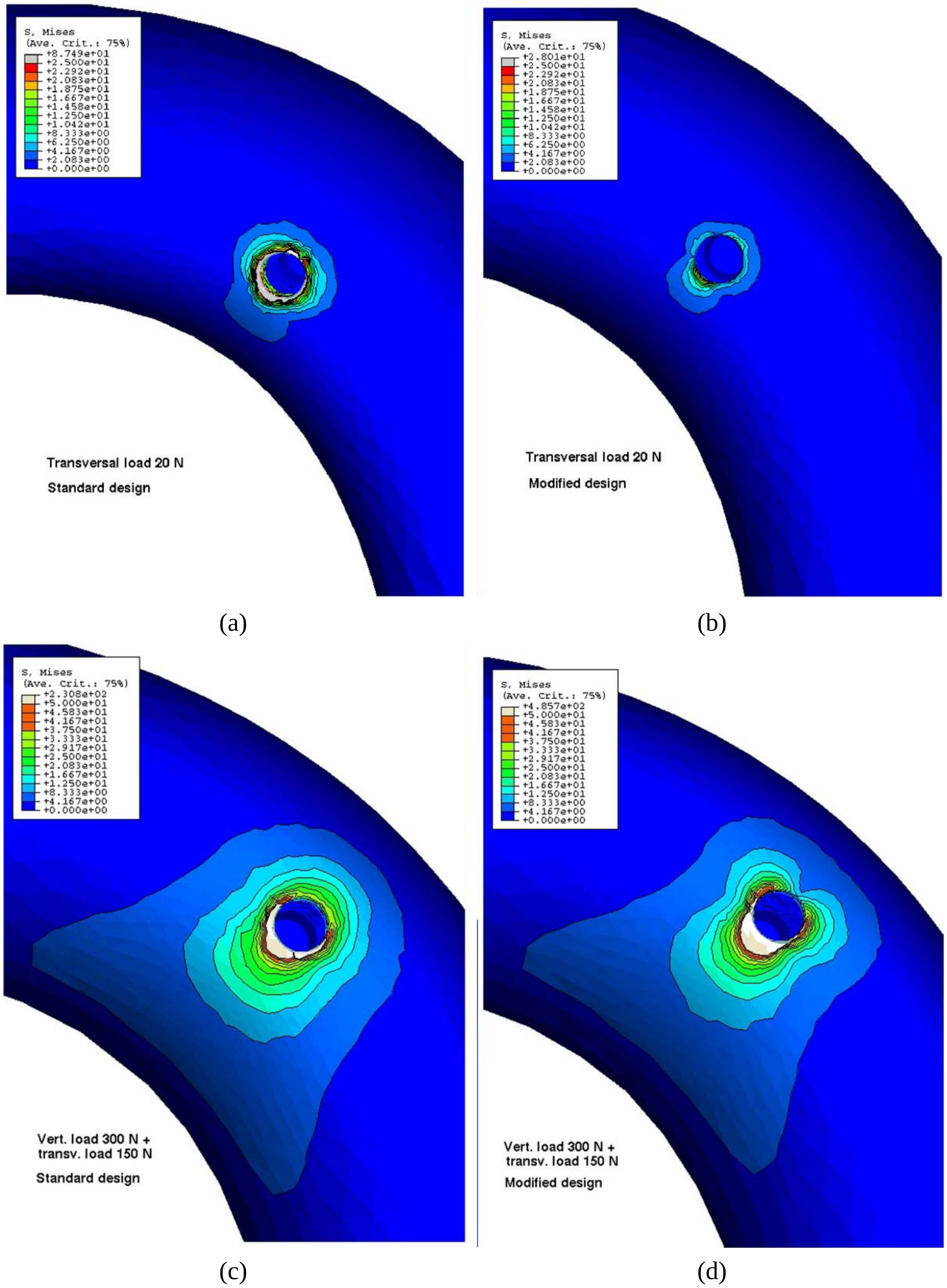

Figure 5 : Comparison of von Mises stress contours in a top view of the jaw bone, for the standard and modified implant designs. (a, c) refer to the standard design; (b, d) to the modified design. (a, b) are for the transversal loading condition; (c, d) for the mixed (axial + transversal) loading condition.

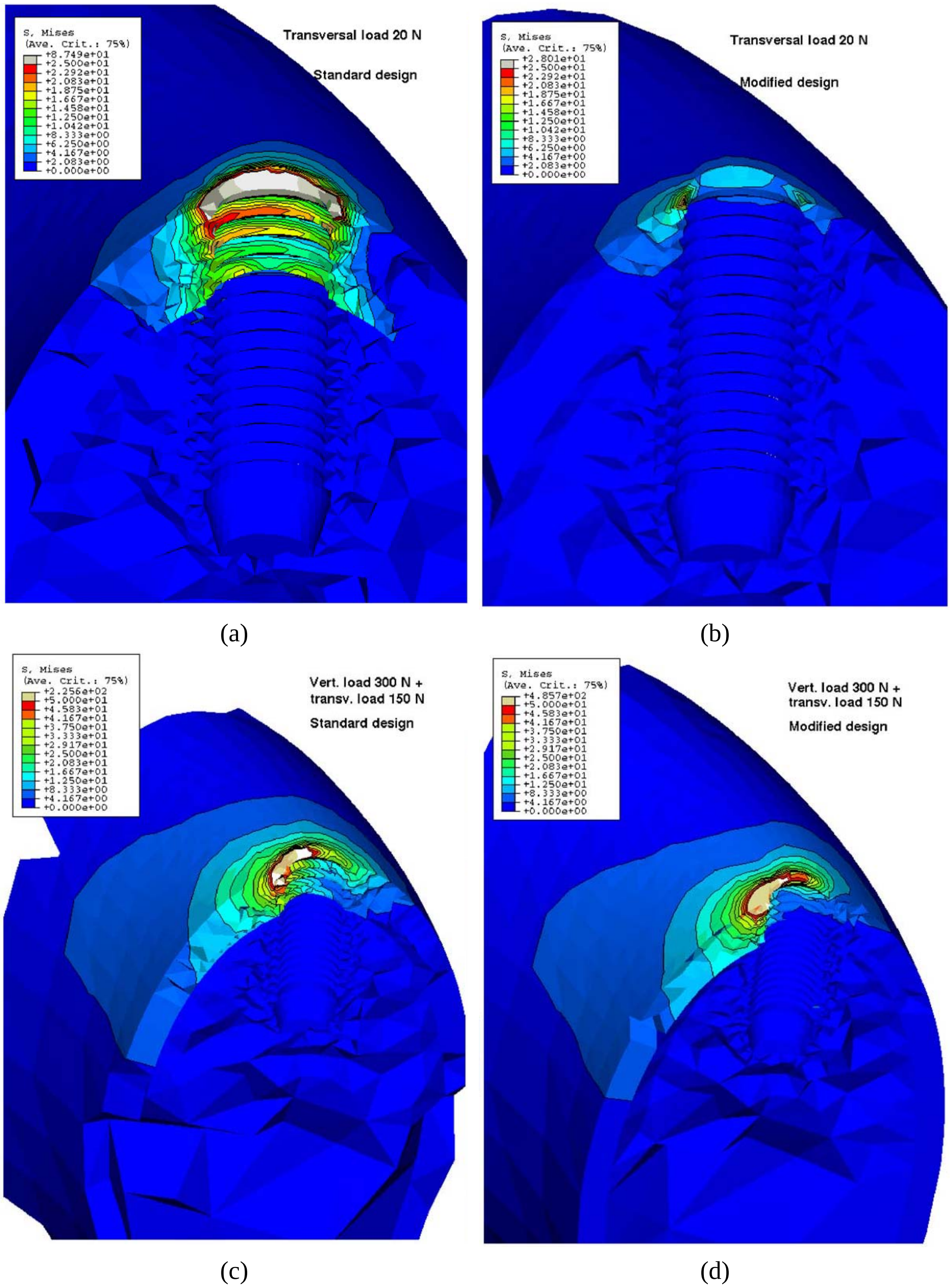

Figure 6 : Comparison of von Mises stress contours in a section of the jaw bone with the sagittal plane, for the standard and modified implant designs. (a, c) refer to the standard design; (b, d) to the modified design. (a, b) are for the transversal loading condition; (c, d) for the mixed (axial + transversal) loading condition.

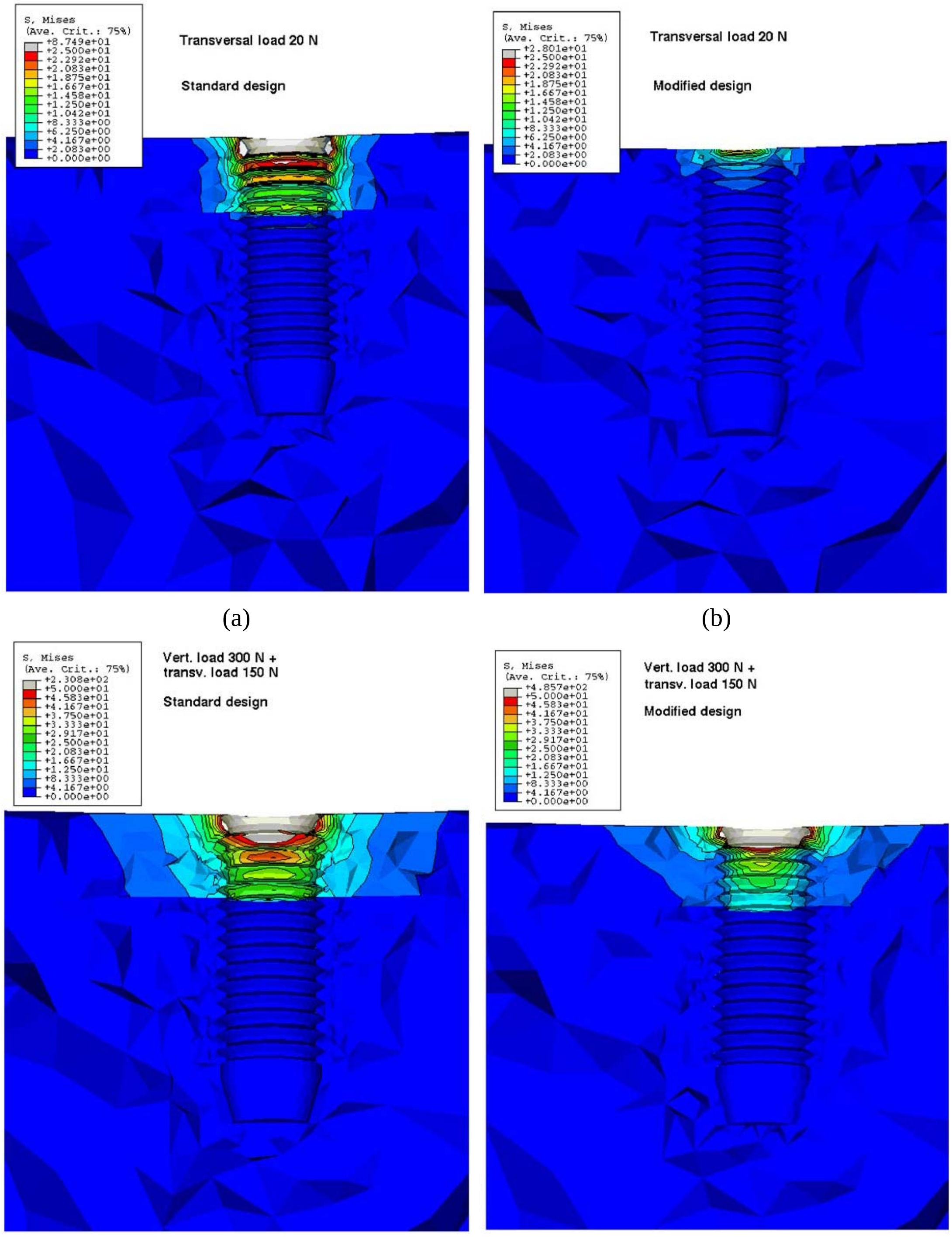

Figure 7 : Comparison of von Mises stress contours in a section of the jaw bone with a vertical plane orthogonal to the sagittal plane, for the standard and modified implant designs. (a, c) refer to the standard design; (b, d) to the modified design. (a, b) are for the transversal loading condition; (c, d) for the mixed (axial + transversal) loading condition

- for the third loading condition the differences between the two designs are still remarkable. The presence of the internal layer causes (i) a different shape of the stress contours in the top surface of the cortical bone (Figures 5c and 5d) and (ii) a doubling of the stress peak in the top threads of the bone, which goes from about 230 to 485 MPa. Also the stress distribution inside the cortical bone is different in the two situations. On the contrary, in the spongious bone the peak von Mises stress goes from about 9 MPa, in the case of the standard implant, to about 5 MPa in the presence of the internal layer; these peaks occur close to the neck of the fixture, but high stress values exist also at its bottom part. In the case of the modified design a further peak of stress occurs, in the spongious bone, also in the middle portion of the threads, an indication of stress redistribution absent in the standard design.

It is interesting to observe that, in the presence of the internal layer, the peak stresses in the bone around the implant tend to *increase*, under high loads, with respect to those computed in the case of a standard design. A similar tendency has been found also in Gei et al. (2002), in the analysis of a healthy tooth-PDL-bone system under axial and transversal loading, comparing models with and without the PDL. The presence of the PDL, simulated as a nonlinear interface whose behavior is that illustrated in Figure 2, seems to cause exactly the same effects, in the jaw bone, as those illustrated here in terms of comparison between two different implant designs.

In the internal layer itself, under the worst loading condition (the rather extreme mixed one) the highest von Mises stress is of about 10 MPa, whereas, under the other loads, the peak stress is of about 3.5 MPa and 4 MPa (axial and transversal, respectively). These values are of the same order of magnitude as those expected in reality within the PDL seen as a continuum material, considering that, according to the results of both Ralph (1982) and Pini (1999), the tensile strength of the PDL is of about 2.5 MPa (see also Figure 2).

Finally, we point out that, as expected, the global mobility curves obtained for the modified implant design match reasonably well, at least qualitatively, the experimental ones for teeth reported in Parfitt (1960) and Brunski (1992).

4 Discussion and Open Problems

We wish to briefly comment about the significance of studying the PDL as a linear elastic medium, as often found in the apparently scarce literature devoted to this issue (see also, for further references, the review paper by Mackerle, 1998). Beside the analyses discussed in the previous Section, we have also tried to study the effect of the internal layer in the modified implant design by treating it as linear elastic, using, for its Young's modulus, the value $E = 50$ MPa. Rees and Jacobsen (1997) suggest this value, for Young's modulus of the PDL, in order to obtain a good match with experimental results for tooth mobility, using a 2–D numerical model, and note that other authors have obtained similarly "good" matches, in a 3–D context, by using $E = 40$ MPa. It is obvious that the "goodness" of such a match, in the presence of a strongly nonlinear behavior, depends on the loading level, and that it is therefore a fictitious goodness.

The results obtained by our linear elastic modeling of the soft layer (not illustrated here for the sake of brevity) are significantly different from those described in the previous Section, obtained adopting a nonlinear model. The largest differences appear for the transversal loading condition, under a "small" load.

The paper by Rees and Jacobsen (1997) quotes experimental results obtained by applying either a transversal load of 2.5 N or an axial load of 20 N. These are low load levels, expected to make the PDL work in a small strain regime, where it has an extremely low stiffness; it is surprising that, in order to match the experimental displacement values, they need such a high value for Young's modulus as 40 or 50 MPa. Indeed, if one computes the tangent longitudinal stiffness of bovine PDL, by numerically differentiating the experimental results of Figure 2a, one can see that the peak tangent modulus value never exceeds 25 MPa, both in tension and in compression (Figure 8). The value $E = 50$ MPa for the PDL is therefore overstiff (not to speak of values like 1750 MPa found in the literature, according to Rees and Jacobsen, 1997); it seems reasonable to ascribe to the 2–D, plane strain numerical model used in Rees and Jacobsen (1997) the cause of this apparent contradiction (unless the bovine PDL is much less stiff than the human one, which seems doubtful). We could not get access to the paper quoted in Rees and Jacobsen (1997), which suggests the value $E = 40$ MPa for the PDL Young's mod-

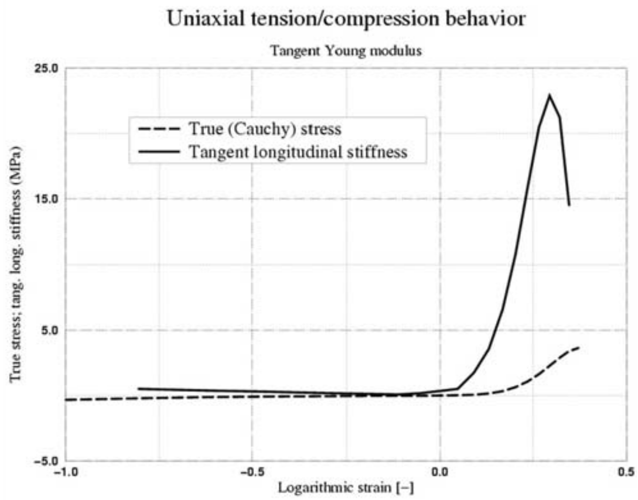

Figure 8 : Tangent longitudinal stiffness for bovine periodontium, computed numerically from the experimental results of Pini (1999).

ulus within the context of a 3–D numerical analysis, and cannot therefore try and understand how this still high value can yield a good match with experimental results.

It thus appears that it is impossible to obtain consistent results by treating the PDL (or its "equivalent" soft layer in our prototype implant design) as a linear elastic medium, for any load range and for any load type. Such a conclusion has been reached also in Gei et al. (2002) with reference to the numerical analysis of the PDL treated as an interface.

Coming back to the mechanical effects of a nonlinear stress-absorbing element, it is easy to understand, also from the results illustrated in Figures 4 to 7, that they are specially important (i) in the presence of prescribed distortions, which cause self-equilibrated stresses; (ii) in the presence of prescribed displacements (not studied here); (iii) for very small loads, when the stiffness of the internal layer is most different from that of the surrounding materials; (iv) in general, in the presence of loading configurations farthest from the statically determined one.

As said in the Introduction, we do not propose, in this paper, a specific material able to reproduce the behavior of PDL and, at the same time, fulfilling all the other requirements necessary in a fixed dental implant (both technological/mechanical and biological/clinical). Here we can only give some indications, at the light of the foreseen requirements, and only from the mechanical viewpoint.

Once agreed about the stiffness requirements — the material should somehow follow the same stress-strain curve as the real PDL — a first necessity is that the internal layer, in a design like that of Figure 1, be made of a material capable to transmit tensile tractions at its interfaces. This is evident both from the results of the analyses illustrated in the previous Section, and from the consideration that, in the case of a multiple implant prosthesis, a single fixture could be subjected to purely axial tensile loads. A second necessity concerns the stress-carrying capacity of this material, which must be appreciable. The maximum von Mises stress computed here in the layer (about 10 MPa, for a quite extreme loading condition, however) agrees well with the values found in Meijer (1995) for the peak stresses in a stress-absorbing layer interposed between fixture and bone (4 to 7 MPa for different stress components). Any material chosen for such a use should be expected to work at these stress levels for a long time, under repeated loading conditions, without suffering any damage.

Other requirements should consider also the damping properties, essential in the definition of the response under dynamic actions. Unfortunately (i) there is absolutely no indication whatsoever, presently, about the damping properties of the PDL and (ii) even in the case these were known, it would be quite a task to match them in an artificial material. This aspect is really far from being tractable at the present state of the research, and we will not add anything more about it here.

A candidate material, for this application, could be some silicone, or siliconic rubber, provided that it guarantees good adhesion properties with titanium. There is no space here to treat this aspect in detail; just to give a first indication, we can observe that the "Silastic" silicone used in aesthetic surgery has a stress-strain curve resembling that of Figure 2a, even if its strength in tension is too low (about 5 MPa), and we do not know much either about its long-term mechanical properties under cyclic loading.

A great deal of attention is obviously required when considering the biological aspects, which may prove even more decisive than the mechanical ones in terms of the choice of the material. Clearly, a key issue from the biological/biocompatibility viewpoint is the interface between the PDL-like layer and the internal/external parts of the fixture, which, at least in the prototype design of Figure 1, is exposed to the intra-oral agents and has to be either accessible for cleaning or fully impregnable to such agents.

Finally, coming back to mechanical/technological aspects, if one wants to retain the internal screw as a connection device between fixture and abutment, one has to study a means of tightening it without straining the soft internal layer, and of guaranteeing, after the tightening, both appropriate mobility, between the osseointegrated and the upper parts of the fixture, and full contact, under all loading conditions, between abutment and fixture.

5 Conclusions

The analyses illustrated in this paper aim to show that

1. a stress-absorbing internal layer in a fixed dental implant can be quite effective in terms both of stress redistribution/absorption and in terms of restoring a mobility close to natural, as long as it is designed and studied in such a way as to behave as closely as possible as the human PDL; no assumption of linear elasticity has meaning, in this respect;

2. in the case of a single, freestanding implant, subjected to external forces, the effect of such an internal layer, in terms of stress diffusion in the bone surrounding the fixture, depends on several aspects, and is expected to be maximum for transversal loads, and minimum, if not negligible at all, under purely axial loads;

3. the effect of a layer with a stiffness much different from that of the surrounding parts is particularly important in the presence of self-equilibrated stresses, and, therefore, is expected to help to significantly reduce both the axial stress in the connection screw caused by its tightening and the self-stresses induced by geometric misfits;

4. even if the results shown here confirm that the presence of an internal PDL-like layer alters significantly the stress state in the jaw bone, it still remains very difficult to assess if such alterations are *good* or *bad* (or insignificant), until a thorough understanding of the real stress state existing around a healthy tooth, under all possible loading conditions, is reached. In this respect refined Finite Element analyses, taking into account both a realistic geometry and a realistic material description, will help; we are still in early stages of the research in this particular field;

5. in any case, under any load, the presence of a PDL-like internal layer would be extremely beneficial in the case of prostheses partially supported by natural teeth, for obvious reasons.

Thus, even without the possibility of quantitatively guaranteeing that by using such a device one might reach an optimal implant design, it appears that the introduction of a PDL-like layer would undoubtedly help improving the basic implant design used nowadays, at least in terms of the mechanical fields inside the bone.

Several research lines in the directions indicated by these conclusions are currently being pursued by this research group. Experiments are under way on pig PDL, in a controlled environment; since the pig denture is more similar to the human one than the bovine one, we expect to obtain more reliable information than what available so far. From the computational viewpoint we have already started to study an efficient way to model the effect of the presence of the PDL on the stress state and the mobility in healthy teeth (Gei et al., 2002), but this approach, based on the use of an interface finite element, does not allow one to go into details of the stress state into the PDL itself. More understanding of this should come from the development of a micromechanical model of the PDL, for instance based on nonlinear beam theory, whose development is currently under way (preliminary results in Perelmuter, 2001).

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The Finite Element code ABAQUS has been run at the Department of Civil Engineering, University of Brescia, under an academic license.

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