Implant–Abutment Interface: Biomechanical Study of Flat Top versus Conical

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ABSTRACT

Background: Overloading has been identified as a primary factor behind dental implant failure. The peak bone stresses normally appear in the marginal bone. The anchorage strength is maximized if the implant is given a design that minimizes the peak bone stress caused by a standardized load. Clinical studies have shown that it is possible to obtain a marginal bone level close to the crest of the implant. Different implant systems make use of different designs of the implant–abutment interface. Different implant–abutment interfaces imply that the functional load is distributed in different ways upon the implant. According to Saint-Venant's principle, this will result in different stress patterns in the marginal bone when this reaches levels close to the implant crest.

Purpose: One aim of the study was to theoretically investigate if a conical implant–abutment interface gives rise to a changed stress pattern in the marginal bone, as compared to a flat top interface, for an axially loaded mandibular titanium implant, the neck of which is provided with retention elements giving effective interlocking with the bone. Further aims were to investigate if the way in which the axial load is distributed on the flat top and on the inner conus respectively affects the stress pattern in the marginal bone. The pertinent stress was considered to be the bone–implant interfacial shear stress. It was assumed that the marginal bone reached the level of the implant–abutment interface.

Method: The investigation was performed by means of axisymmetric finite element analysis.

Results: The conical implant–abutment interface of the type studied brought about a decrease in the peak bone–implant interfacial shear stress as compared to the flat top interface of the type studied. This peak interfacial shear stress was located at the top marginal bone for the flat top implant–abutment interface whereas it was located more apically in the bone for the conical implant–abutment interface. The way in which the axial load was distributed on the flat top and on the inner conus respectively affected the peak interfacial shear stress level.

Conclusion: The design of the implant–abutment interface has a profound effect upon the stress state in the marginal bone when this reaches the level of this interface. The implant with the conical interface can theoretically resist a larger axial load than the implant with the flat top interface.

KEY WORDS: anchorage strength, bone stress, implant-abutment interface

Dental implants are tiny structures that are subject to considerable loads. The clinical results of treatment with maxillary implants are regularly inferior to the results of mandibular implants.[1] The lower jaw normally presents higher quality bone than the upper jaw.[1] The success rate seems to be correlated to the amount and the quality of bone. Longer implants have been observed to score better than shorter ones.[2,3] Overloading has been identified as the main etiologic factor behind loss of the Brånemark implant.[2] Put together, this suggests that overloading is a primary factor for loss of dental implants.[4]

In dental implant design, an important issue should consequently be to find the implant shape that maximizes the anchorage strength in bone. If bone is subject to extreme stress, it will be resorbed.[5] According to general engineering principles, the anchorage strength is maximized if the implant is given a design that minimizes the peak bone stress caused by a standardized load.[4] The smaller the peak stress caused by a standardized load, the greater the load required to trigger bone fracture or bone resorption. In a number of finite element studies, it has been found that the peak stress in

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the bone supporting a dental implant appears in the crestal region close to the level where the implant starts to attach to the bone.[4,6–9]

Based upon finite element analysis, Stoiber[7] and Mailath et al[8] argued for a smooth implant neck to allow sliding motion between implant and bone. This would allow the cortical bone to resist horizontal load components whereas vertical load components are resisted by the cancellous bone. The rationale for this recommendation was that the peak stress caused by horizontal load components should be spatially separated from the peak stresses caused by vertical load components. However, in a finite element study of an axially loaded dental implant with a flat top design, Hansson[4] obtained a considerably decreased peak interfacial shear stress for a high bone attachment level as compared to a low attachment level.

Adell et al[10] observed a mean marginal bone loss of 1.5 mm during healing and the first year after bridge connection for the Brånemark implant with its flat top implant–abutment interface. In a clinical study comprising four different screw-shaped dental implants provided with a smooth neck of different lengths, Jung et al[11] found that the marginal bone stabilized at the level of the first thread. The above studies indicate a marginal bone level at some distance from the implant–abutment interface.

In an animal study with loaded porous-coated dental implants provided with a machined neck of two different heights, Al-Sayyed et al[12] found that the marginal bone resorption stopped at the junction between the machined surface and the porous surface. It was suggested that the marginal bone resorption was caused by disuse atrophy in accordance with Wolff's law.[13] Palmer et al[14] observed unusually high marginal bone levels using an implant characterized by a conical implant–abutment interface and a conical neck that is provided with a thread of small dimensions and roughened by means of blasting with titanium dioxide. The load from the abutment is transmitted to an inner conus in the implant. The marginal bone stabilized close to the level of the implant–abutment interface. It has been suggested that the high marginal bone level was due to the presence of retention elements providing for good interlocking with the bone.[4] Other studies have confirmed the marginal bone maintenance around this implant.[15–17]

This study addresses the question of whether the design of the implant–abutment interface affects the stress distribution in the marginal bone. According to Saint-Venant's principle,[18] a change in the distribution of the load on the end of a structure, without change of the resultant, alters the stresses only near the end. For this reason, it can be concluded that the design of the implant–abutment interface does not affect the stresses in the marginal bone for implants where the marginal bone level is located at some distance from the implant–abutment interface. For an implant of the type studied by Palmer et al[14] however, with a marginal bone level close to the implant–abutment interface, it can be suspected that the design of this interface affects the stresses in the marginal bone.

One aim of the present study was to investigate, by means of finite element analysis, if a conical implant–abutment interface gives rise to a changed stress pattern in the surrounding bone, as compared to a flat top interface, for an axially loaded mandibular titanium implant, the neck of which is provided with retention elements giving effective interlocking with the bone (Figure 1). Further aims were to investigate if the way in which the axial load is distributed on the flat top and on the inner conus respectively affects the stress pattern in the marginal bone. It was assumed that the marginal bone reached the level of the implant–abutment interface. The overall implant design chosen was close to that of the implant depicted in Figure 2.

METHOD

The investigation was performed by means of stress analysis using the finite element method.[19] The finite element program used was NISA (Engineering Mechanics Research Corporation, Troy, Michigan, USA). An axisymmetric model (Figure 3) was used, the dimensions and elastic constants of which were in a previous study[4,20] derived from a three-dimensional model of the mandible. The outline of the procedure to obtain the model is given below. A more detailed description of this procedure is given in Hansson.[4,20] A three-dimensional mandibular model was subjected to a standardized vertical load at the center of the upper cortical shell. The elastic constants of this model were obtained from Carter,[21] Hart et al,[22] and Turner.[23] The difference in vertical displacement between the center of the upper and the lower cortical shell was calculated. The dimensions of the axisymmetric model were adjusted in such a way that when subjected to the same vertical load at the central part of the upper cortical shell, the same difference in

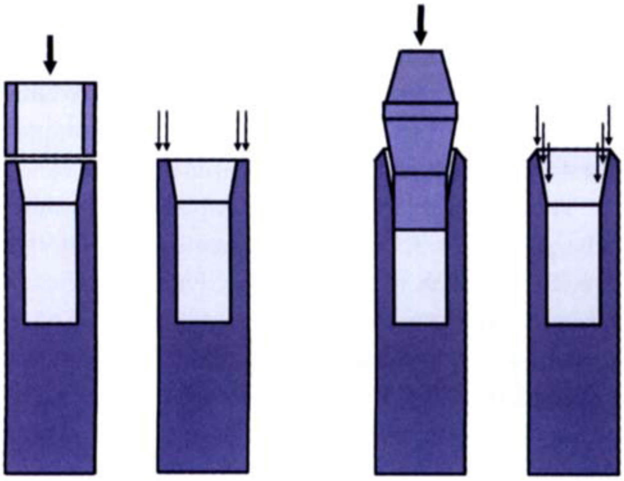

Figure 1. To the left, an implant with a flat top fixture–abutment interface. The load from the abutment is transferred to a peripheral horizontal surface on top of the fixture. To the right, an implant with a conical fixture–abutment interface. The load from the abutment is transferred to an inner conical surface on the fixture.

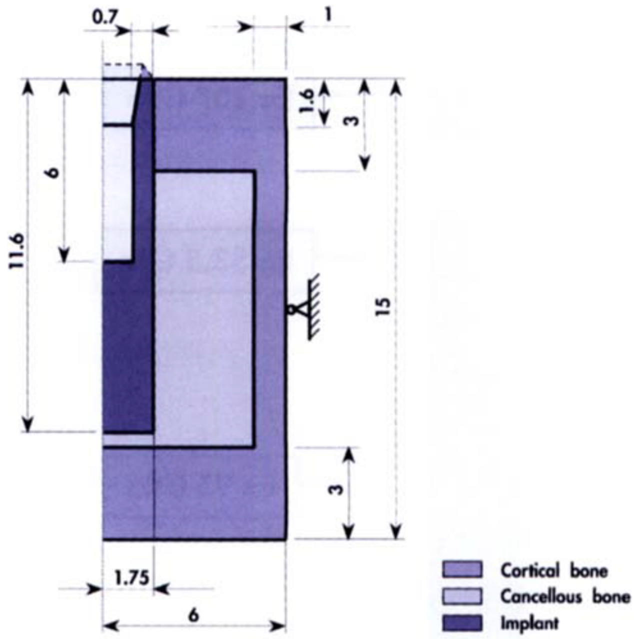

Figure 3. The axisymmetric finite element model (measures in mm).

vertical displacement between the central parts of the upper and lower cortical shells arose.

The bone and the titanium were assumed to be isotropic and linearly elastic with elastic constants[4] according to Table 1. The implant was modelled as an 11.6- to 12-mm-long (the length depending upon the type of implant–abutment interface applied), 3.5-mm-wide cylinder provided with a bore (see Figures 1, 3). One important parameter determining the mechanical behavior of an implant is its axial stiffness.[4] The axial stiffness is the product of the cross-sectional material area and the modulus of elasticity. If the implant had been provided with a thread, the cross-sectional material area would have been reduced. In order to, in a mechanical sense, simulate a thread at the apical 10 mm of the implant, the modulus of elasticity of this part of the implant was decreased (Figure 4). By this measure, the modelled cylindrical implant exhibited an axial stiffness close to that of a screw-shaped implant provided with a cylindrical neck.

Eight-node elements were used exclusively. Two different implant–abutment interfaces were modelled: a flat top interface (Figures 1, 5) and a conical interface (Figures 1, 6). Two different load cases were used for the flat top interface: Flat 1-7, where a standard load of 1000 N was equally distributed upon all of the nodes on the flat top, and Flat 4-7, where the same load was equally distributed upon the four lateral implant nodes on the flat top (Figures 5, 7). For the conical interface, the following designation of different load cases were used. Conical 1 means that a load 1000 N was located on node 1. Conical 1-n means that this load was equally distributed upon the nodes 1 through n (Figures 6, 8). The interfacial shear stress and principal stress 1, principal stress 2, and principal stress 3[18] in the marginal bone were calculated.

Figure 2. An implant that resembles the modelled implant with the conical implant–abutment interface (Fixture Microthread™, Astra Tech AB, Mölndal, Sweden). This implant has a longer "cylindrical portion."

TABLE 1. Elastic Constants Used in the Finite Element Model

Titanium	E = 107 GPa	ν = 0.34
Cortical bone	E = 15 Gpa	ν = 0.3
Cancellous bone	E = 456 Mpa	ν = 0.2

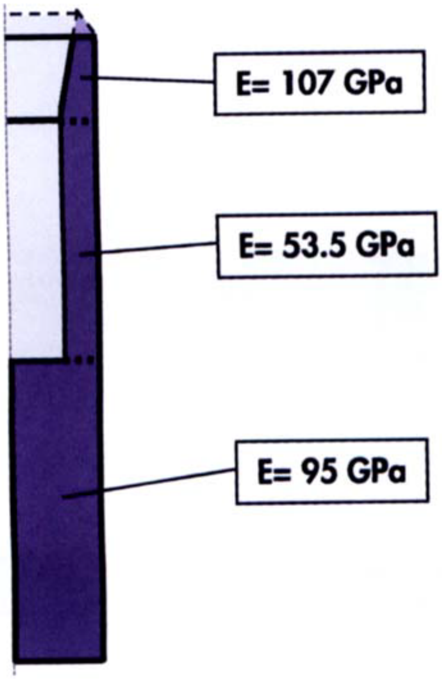

Figure 4. The moduli of elasticity in the different parts of the cylindrical implant model. Poissons ratio was assumed to be 0.34 for all parts of the implant.

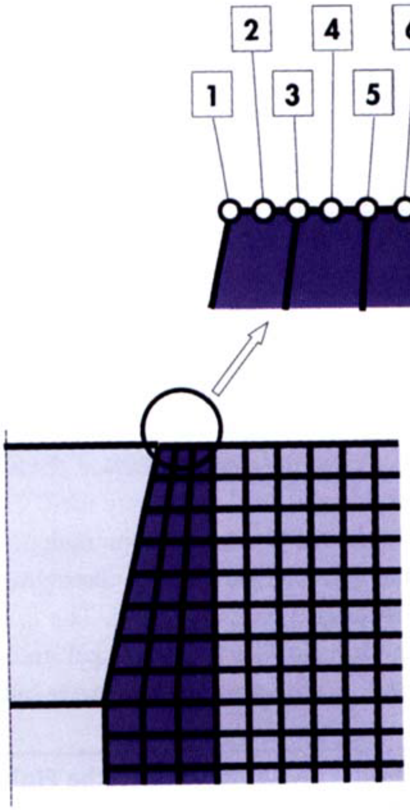

Figure 5. The nodes on the top of the flat top implant upon which loads were applied. In load case Flat 4-7, the standardized load was equally distributed upon the nodes 4-7, whereas in load case Flat 1-7, the standardized load was equally distributed upon the nodes 1-7.

At an arbitrary point in a loaded solid, three orthogonal planes exist at which the shear stresses are zero. The tensile and/or compressive stresses at these planes are called the principal stresses. The extreme tensile and compressive stresses at any point are always principal stresses.

The cylindrical surface of the implant was assumed to be provided with retention elements giving effective interlocking with regard to shear. Possible adhesion between titanium and bone was neglected. Consequently, the cylindrical bone–implant interface was assumed to resist shear stresses and compressive stresses but not tensile stresses. The feature of resisting compressive stresses but not tensile stresses was modelled by connecting and disconnecting, in a direction perpendicular to the interface, the interfacial implant nodes and bone nodes (in couples) in an iterative procedure with repeated FEM calculations until the intended interfacial conditions were mimicked. The details of this procedure were as follows: (1) All interfacial implant and bone nodes were connected in a horizontal direction. A FEM calculation was made. (2) The results of the FEM calculation were analyzed. Where horizontal interfacial tensile stresses occurred, the interfacial implant and bone nodes were disconnected in a horizontal direction. Another FEM calculation was made. The gaps arising between implant and bone were normally in the submicron range. (3) If horizontal interfacial tensile stresses still occurred, the interfacial implant and bone nodes in question were disconnected in a horizontal direction. If implant and bone had penetrated into each other (where the interfacial implant and bone nodes were disconnected), the interfacial implant and bone nodes were reconnected in a hori-

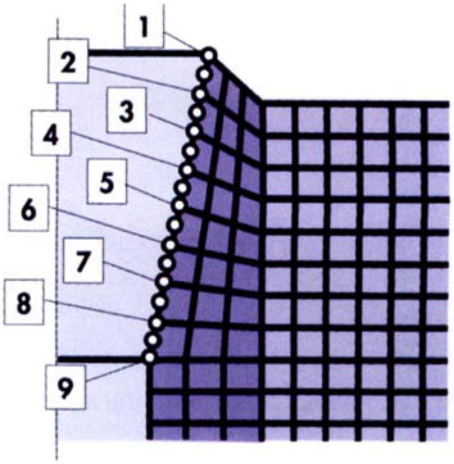

Figure 6. The nodes on the inner conus of the implant with the conical fixture–abutment interface upon which loads were applied.

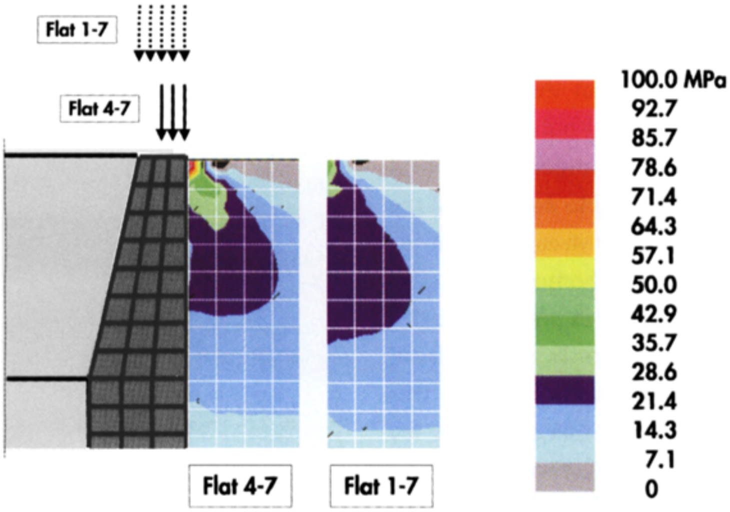

Figure 7. Bone-implant interfacial shear stress contours for the load cases Flat 4-7 and Flat 1-7 for the flat top implant.

zontal direction. Another FEM calculation was made. This procedure was continued until no horizontal interfacial tensile stresses remained and bone at no location had penetrated into the implant. Interlocking with regard to shear was modelled by connecting (in couples) the interfacial implant nodes and bone nodes in a vertical direction. Horizontal and vertical connections were specified independently of each other. The number of elements in the model was 2025.

RESULTS

The peak interfacial shear stress around the implant neck for the different load cases is shown in Table 2. The contours for the interfacial shear stress are shown in Figures 7 and 8. The peak interfacial shear stress for the load cases Conical 1 and Conical 1-n were smaller than for the load cases Flat 4-7 and Flat 1-7. For the load cases Flat 4-7 and Flat 1-7, the peak interfacial shear stress was located at the top marginal bone, whereas it was located deeper in the bone for the load cases Conical 1 and Conical 1-n.

The principal stress 1, principal stress 2, and principal stress 3 at the top marginal bone for the different load cases are given in Table 3. For all load cases, the peak principal stress 1 was located down in the bone. For principal stress 2 and principal stress 3, however, the values at the top margin represent the peak values.

DISCUSSION

The peak principal stresses (see Table 3) in the interfacial bone were substantially higher than the peak interfacial shear stresses (see Table 2). However, these peak principal stresses were compressive and, considering the properties

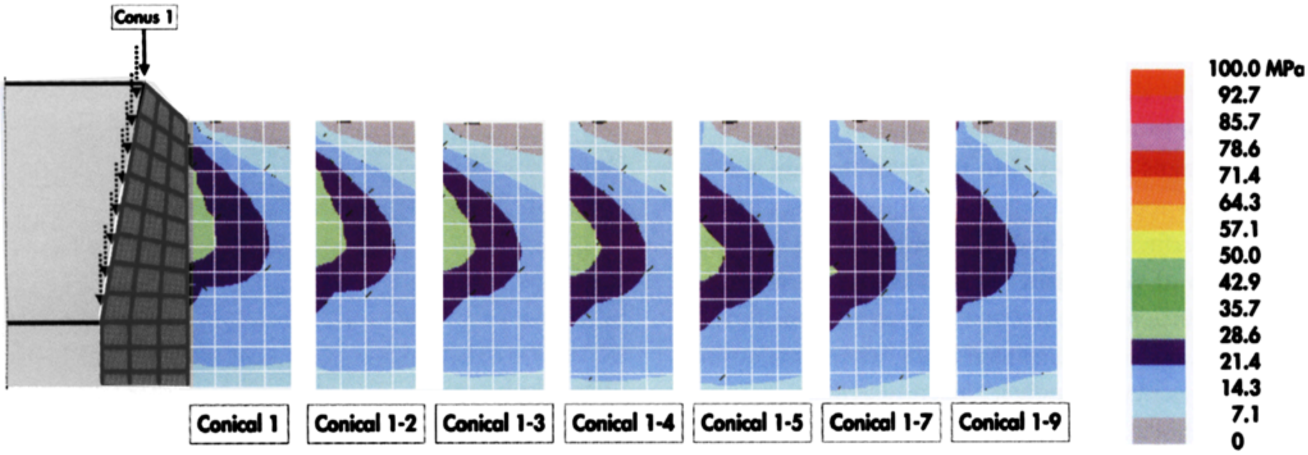

Figure 8. Bone-implant interfacial shear stress contours for the different load cases for the implant with the conical fixture–abutment interface.

of human cortical bone, the strength in compression is about 2.0 to 2.8 times the strength in shear.[24] Furthermore, it seems natural to assume that the strength in compression of the interfacial cortical bone is not substantially different from that of homogenous cortical bone. On the contrary, it seems natural to assume that the cortical bone–implant interfacial shear strength is considerably inferior to the shear strength of homogenous cortical bone, which has also been verified in push-out tests[25] and analyzed theoretically.[26] Consequently, the interfacial shear stress is suggested to be the crucial stress in this context.[4] The principal stresses in the marginal bone are only considered to give supplementary information.

Theoretically, the interfacial shear stress shows exactly how an axial load upon a cylindrical implant is transferred into the bone. A high value of the interfacial shear stress implies an abrupt load transfer, which is considered to be unfavorable, whereas a moderate interfacial shear stress signifies a gradual load transfer into the bone. When the interfacial shear stress exceeds the interfacial shear strength, bone fracture and relative movements or bone resorption can be expected to occur. High interfacial shear stresses have been associated with interfacial slip and interfacial failure for total hip arthroplasties.[27]

It might be argued that it is a serious limitation that only the effect of axial loads was investigated in this study. Consider an implant-supported fixed bridge subject to vertical and horizontal loads. The vertical and horizontal loads will give rise to vertical and horizontal forces upon the implants. Furthermore, the loads will give rise to bending moments in the bridge. If the bridge has sufficient stiffness and if the implants are not located along a straight line, these bending moments will as a first approximation be resisted by axial forces in the implants.[28,29] If these conditions are not fulfilled, the implants will also be subjected to bending moments. Horizontal loads and bending moments upon implants of the designs considered in the present study will not give rise to appreciable interfacial shear stresses but will cause interfacial normal stresses as well as principal stresses and von Mises stresses in the surrounding bone.[4,7] Since interfacial shear stresses were the focus of this study, it can be concluded that it was no major limitation to restrict the study to effects of axial loads.

The choice of 1000 N as the magnitude of the standardized axial load might appear arbitrary. However, all materials were assumed to be linearly elastic, which implies that there is a direct proportionality between loads and stresses provided that the deformations and displacements are not so big that the geometry of the structure is changed. The only nonlinear feature introduced into the model was the iterative connection and disconnection of the interfacial implant nodes and bone nodes in order to simulate resistance to compressive but not to tensile stresses. A check with a load of 100 N showed that the relative magnitude of the peak interfacial shear stresses for the different designs and load applications subsisted.

It could be argued that the element mesh used is rather crude. However, it was not the absolute values of the peak stresses that were sought but their relative values and locations, and the model used gives a fairly true

TABLE 2. Peak Interfacial Shear Stress for the Different Load Cases

Load Case	Peak Interfacial Shear Stress (MPa)
Flat 4–7	100.4
Flat 1–7	44.4
Conical 1	30.6
Conical 1–2	31.6
Conical 1–3	32.1
Conical 1–4	32.3
Conical 1–5	31.7
Conical 1–6	31.0
Conical 1–7	29.7
Conical 1–8	28.1
Conical 1–9	26.9

TABLE 3. Principal Stress 1, Principal Stress 2, and Principal Stress 3 at the Top Margin for the Different Load Cases

Load Case	Principal Stress 1 (MPa)	Principal Stress 2 (MPa)	Principal Stress 3 (MPa)
Flat 4–7	−57.1	−117.7	−277.7
Flat 1–7	−32.4	−80.4	−149.8
Conical 1	−12.6	−57.3	−105.0
Conical 1–2	−9.1	−50.8	−74.7
Conical 1–3	−8.5	−50.1	−69.1
Conical 1–4	−10.1	−54.0	−67.6
Conical 1–5	−11.9	−58.1	−76.8
Conical 1–6	−13.7	−61.6	−85.5
Conical 1–7	−15.2	−64.3	−92.8
Conical 1–8	−16.4	−66.3	−98.7
Conical 1–9	−17.4	−67.8	−103.3

picture in these respects. The dimensions of the axisymmetric model (see Figure 3) used were in a previous study[4,20] derived from a three-dimensional model of the mandible. This three-dimensional model had an idealized cross-section, which probably does not often show up in clinical practice. However, Hansson[4] found that different modifications of the axisymmetric bone model did not significantly alter the effect of some implant design changes upon the peak interfacial shear stress. The modifications introduced were (1) upper and lower cortical thickness = 2.5 mm, (2) upper and lower cortical thickness = 4 mm, (3) lateral extension (radius) = 4 mm, (4) lateral extension (radius) = 5 mm, (5) rounded lateral corners with an external cortical radius of curvature of 3 mm and an internal cortical radius of curvature of 1 mm, (6) moduli of elasticity of cortical and cancellous bone reduced by 50%, and (7) moduli of elasticity of cortical and cancellous bone increased by 100%. This suggests that the results obtained in this study have a certain universal validity.

It was suggested that the interfacial shear stresses are the crucial stresses in this context. However, it can also be of interest to discuss the principal stresses in the interfacial bone. Then it is also necessary to include the effect of horizontal loads and bending moments upon the implant. In a finite element study, Stoiber[7] found that when a cylindrical implant is subjected to a horizontal load, the peak interfacial principal bone stresses land at the top of the marginal bone. The horizontal load was applied at a certain level above the bone. A horizontal load at one level is interchangeable with the same horizontal load at another level and a bending moment. Thus, it can be suspected and is intuitively understood that the peak stresses resulting from bending moments also land at the top marginal bone. The mandible can be regarded as a bent beam. The masticatory forces give rise to bending moments in this beam. For the sake of simplicity, consider a straight beam subjected to bending moments according to Figure 9. According to classical beam theory,[30] these bending moments give rise to compressive stresses above the neutral surface and tensile stresses below the neutral surface. The peak compressive stress is located at the very top of the beam and the peak tensile stress at the very bottom. If the bending moment is reversed, the stress pattern will be reversed. The bending moment vector resultants acting upon the mandible during function and parafunction may have different directions. But irrespective of the direction of these bending moment vector resultants, the peak stresses resulting from them always land at the surface of the mandible. The deeper in the mandible, the more insignificant are these stresses. The application of the axial load at the conical interface brought about a substantial decrease in the peak principal stresses at the marginal bone (see Table 3) as compared to an application of the load at the flat top. In a situation with both an axial and a horizontal load on the implant, the bone stresses caused by the axial load component will be added to those caused by the horizontal load component and by the bending moments in the mandible. There is no reason to believe that the magnitude of the bending moments in the mandible would be affected by the design of the implant–abutment interface, nor is there any reason to believe that the conical and the flat top interface would differ substantially with respect to bone stresses resulting from horizontal loads. Consequently, it can be suspected that the conical implant–abutment interface studied compares favorably with the flat top interface also where principal interfacial bone stresses are concerned. A detailed analysis of these principal stresses would, however, require a much more sophisticated full three-dimensional finite element model.

The application of the load at the conical interface brought about a substantial decrease in the peak interfacial shear stress as compared to an application of the load at the flat top. The smaller the peak interfacial shear stress caused by the standard load, the bigger the load required to trigger bone resorption or to achieve bone fracture and interfacial slip. This means that the implant with the conical interface theoretically can resist a bigger axial load than the implant with the flat top before bone resorption is triggered due to high

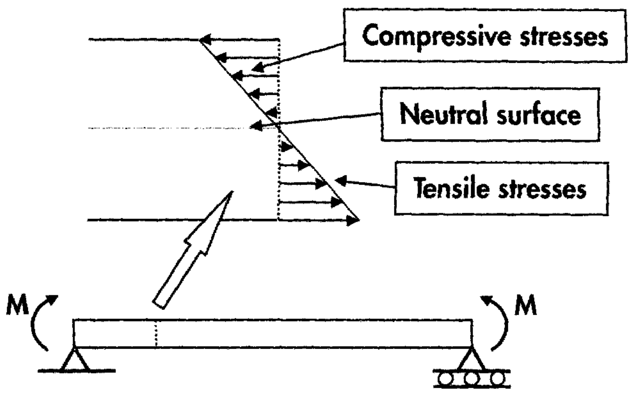

Figure 9. The stresses in a straight beam subjected to a uniform bending moment.

stress concentrations. Furthermore, the application of the load at the conical interface moved the peak interfacial shear stress from the level where the bone starts to get attached to the implant further down into the bone. The further down in the inner conus the load was distributed, the further down the peak interfacial shear stress landed. According to the above, the deeper down in the bone, the smaller are the stresses that can be expected to result from horizontal loads or bending moments on the implant and from the mandible functioning as a beam. It is suggested that it is favorable to spatially separate the peak interfacial shear stresses caused by axial load components on the implant from the peak stresses caused by horizontal load components and by the mandible functioning as a beam. This is in essence what was suggested by Stoiber[7] and Mailath et al.[8] Consequently, it appears to be favorable to have the axial load distributed deeply in the conus. Thus, the conical implant–abutment interface appears to some extent to solve the previously mentioned problem posed by Stoiber[7] and Mailath et al.[8] With a conical implant–abutment interface of the type studied, the location of the peak interfacial shear stress caused by an axial load appears to be spatially separated from the peak stress caused by a horizontal load and by the mandible functioning as a beam.

Also, with the flat top design, the location of the load affected the stresses in the marginal bone. The results suggest that the more central the location of the load, the more favorable is the stress distribution in the marginal bone.

As regards the stresses in the neck region, it can be assumed that the results do not apply only to a perfectly cylindrical implant neck with a rough surface, achieving good interlocking,[26] but also with good approximation to a cylindrical implant neck equipped with retention elements such as threads (see Figure 2), provided that the dimensions of these retention elements are small.[31] A thread of small dimensions as well as the asperities of a rough surface can be regarded as irregularities that are able to resist shear forces. As long as perfect interlocking between implant and bone is prevailing apical to the neck region, the interfacial shear stresses in the neck region do not depend upon whether this interlocking is achieved by means of a rough surface or a macroscopic thread. What matters for the interfacial shear stresses in the neck region is the axial stiffness of the implant apically to the neck.[4] This means that with good approximation the results apply equally to an implant with a thread of small dimensions at the neck and a standard thread apically to the neck.

The design of the implant–abutment interface appears to have a profound effect upon the stress situation in the marginal bone when this reaches the level of this interface. It is suggested that the implant–abutment interface be designed in such a way that the peak bone–implant interfacial shear stress caused by an axial load is reduced and does not land at the very attachment level where the implant starts to interlock with the bone, but deeper down. A flat top interface of the design in question seems to give rise to unnecessarily high peak bone-implant interfacial shear stresses, with an unfavorable location of these peak stresses. In contrast, a conical implant–abutment interface appears to be a design that gives rise to moderate peak interfacial shear stresses and a more favorable peak stress location.

In more general terms, the results suggest that the more central and deeper down in the implant is the point of application of the axial load, the more favorable is the stress distribution in the marginal bone.

CONCLUSIONS

For a dental implant, the neck of which is provided with effective retention elements and where the marginal bone reaches the level of the implant–abutment interface,

- The design of this interface has a profound effect upon the stress state in the marginal bone.
- A conical implant–abutment interface of the type studied brings about a decrease in the peak bone-implant interfacial shear stress resulting from an axial load as compared to a flat top interface of the type studied. This means that an implant with the conical interface theoretically can resist a bigger axial load than the same implant with the flat top interface, before bone resorption is triggered due to high stress concentrations.
- The peak bone–implant interfacial shear stress resulting from an axial load is located at the top marginal bone for a flat top implant–abutment interface of the type studied, whereas it lands further down in the bone for a conical implant–abutment interface of the type studied. The further down in the inner conus the axial load is distributed, the further down the peak interfacial shear stress lands. This means that with the

conical implant–abutment interface, the peak interfacial shear stresses caused by axial load components upon the implant are spatially separated from the peak stresses caused by horizontal load components and by the mandible functioning as a beam.

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