Finite Element Analysis of Stress-Breaking Attachments on Maxillary Implant-Retained Overdentures

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Purpose: The purpose of this study was to examine the effect of stress-breaking attachments at the connections between maxillary palateless overdentures and implants. ***Materials and Methods:*** Three-dimensional finite element models were used to reproduce an edentulous human maxilla with an implant-retained overdenture. Two-implant models (in the canine tooth positions on both sides) and four-implant models (in the canine and second premolar tooth positions on both sides) were examined. Stress-breaking material connecting the implants and denture was included around each abutment. Axial loads of 100 N were applied to the occlusal surface at the left first molar tooth positions. In each model, the influence of the stress-breaking attachments was compared by changing the elastic modulus from 1 to 3,000 MPa and the thickness of the stress-breaking material from 1 to 3 mm. Maximum stress at the implant-bone interface and stress at the cortical bone surface just under the loading point were calculated. ***Results:*** In all models, maximum stress at the implant-bone interface with implants located in the canine tooth position was generated at the peri-implant bone on the loading side. As the elastic modulus of the stress-breaking materials increased, the stress increased at the implant-bone interface and decreased at the cortical bone surface. Moreover, stress at the implant-bone interface with 3-mm-thick stress-breaking material was smaller than that with 1-mm-thick material. ***Conclusion:*** Within the limitations of this experiment, stress generated at the implant-bone interface could be controlled by altering the elastic modulus and thickness of the stress-breaking materials. *Int J Prosthodont 2007;20:193–198.*

To provide adequate support, stability, and retention, maxillary complete dentures generally have palatal coverage.[1] Several studies have evaluated the physical response to different amounts of prosthetic palatal coverage,[2–4] suggesting that to maintain oral sensation and function it is actually advantageous not to include palatal coverage. However, because of continuing reduction of the residual ridges during protracted wearing of dentures,[5,6] it is often difficult to maintain retention and stability of palateless overdentures. In such cases, an implant-retained overdenture is one treatment option.[7–9]

In some cases, the residual ridges allow numerous implants to be placed, but to reduce surgical, psychologic, and economic stresses,[9] a minimal number of implant placements is sometimes selected. On the other hand, in patients with advanced residual ridge resorption, the opportunities for implant placement are limited. To improve this situation, extensive surgical reconstructive procedures are often needed before implant placement; however, these procedures are often difficult because of the stress placed on the patient. It is therefore necessary to consider the use of an efficient minimal number of implants.

Because the palatal support area of palateless dentures is less than that of dentures with palatal coverage, the stress on implants of palateless overdentures

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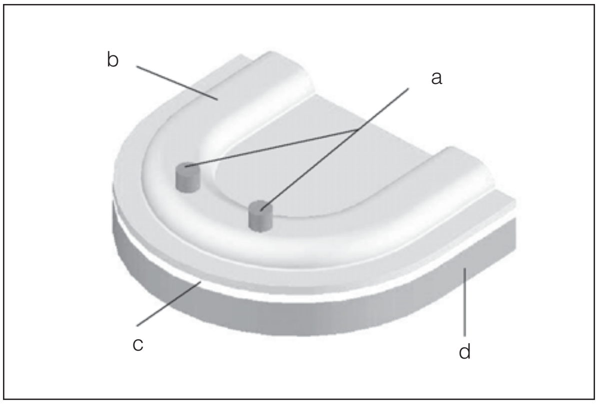

Fig 1 Model maxilla. a: Implant; b: mucosa; c: cortical bone; d: cancellous bone.

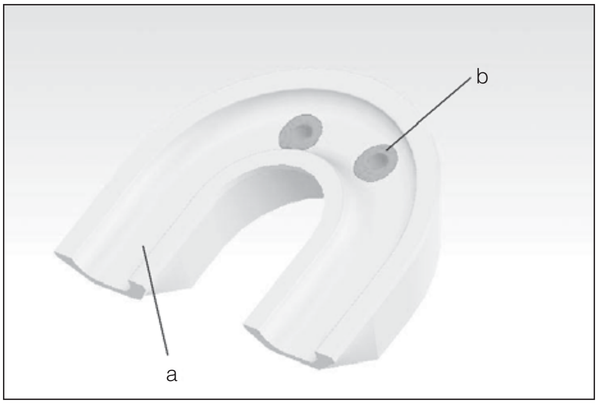

Fig 2 Superstructure of the model. a: Denture; b: resilient material.

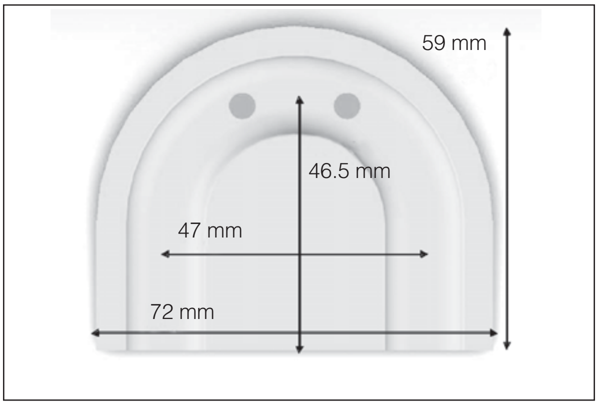

Fig 3 Dimensions of the ridge.

is larger.[9] Only a minimal number of implants is likely possible at the implant-bone interface, where stress is concentrated when the denture is displaced by occlusal contact. Excessive stress concentration causes bone resorption and damages implant osseointegration.[10] To avoid stress concentration, use of stress-breaking materials at the connections between overdentures and implants is considered effective. However, this may alter the denture displacement, increasing stress and aggravating resorption at the residual ridges. Currently, the ability of stress-breaking materials to reduce stress on the implants and ridges remains almost unknown.

Stress dynamic analyses of implants have been conducted using photoelasticity, strain-gauge measurements, and finite element analysis.[11] Photoelasticity provides good qualitative information on the overall location and concentration of stresses but produces limited quantitative information. Strain-gauge measurements only provide data regarding strain at the gauge, with the influence of gauge size remaining undetermined. Finite element analysis can simulate stress dynamics using a computer-created model to calculate stress, strain, and displacement. Such analysis has the advantage of allowing several conditions to be changed easily[12–15] and also allows measurement of stress distribution around implants at optional points that are difficult to examine clinically. Consequently, this approach has been widely used for analyses of stress dynamics.

In the present study, the aim was to analyze the effect of stress-breaking attachments on stress distribution with maxillary palateless implant-retained overdentures using finite element analysis. The purpose was to reveal the effect of such attachments at the connections between the overdentures and implants on stresses at the implant-bone interface under loading during mastication and on the cortical bone of the ridge just under the loading point.

Materials and Methods

Three-Dimensional Finite Element Models

Three-dimensional (3D) finite element models were reproduced using the 3D CAD program (SolidWorks 2005, SolidWorks) to represent an edentulous human maxilla. These models included 2-mm-thick cortical bone and mucosa, cancellous bone, implants, stress-breaking attachments, and dentures (Figs 1 to 3).[12,16,17] The residual ridge was represented by moderate resorption.[18] A 2-implant model (implants were located in the canine tooth positions on both sides) and 4-implant model (implants were located in the canine and second premolar tooth positions on both sides) were examined. The types of implants chosen for modeling were 3.3-mm diameter, 10-mm titanium implants in the canine positions and 4.1-mm diameter, 10-mm titanium implants in the premolar positions (ITI, Straumann). Although implants more than 4.1 mm in diameter and longer than 10 mm are recommended,[19] a narrow diameter is preferable in the canine position because the ridge is often narrow. Therefore, a diameter of 3.3 mm was chosen. Implant abutments were cylindric with a height of 4 mm from the mucosal surface and a diameter of 4.8 mm. Stress-

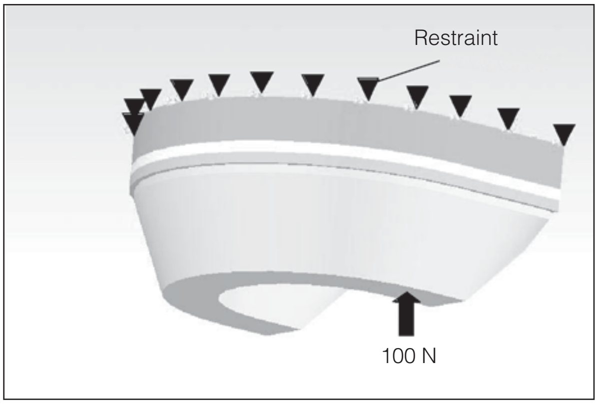

Fig 4 Load applied and restraint conditions.

Table 1 Material Properties of the Dentures, Mucosa, Cortical Bone, Cancellous Bone, Implants, and Stress-Breaking Material

	Elastic modulus (MPa)	Poisson ratio
Denture (acrylic resin)	3,000	0.35
Mucosa	1	0.37
Cortical bone	13,700	0.30
Cancellous bone	1,370	0.30
Implant (titanium)	103,400	0.35
Stress-breaking material	1, 15, 30, 250, 500, 1,000, 2,000, 3,000	0.30

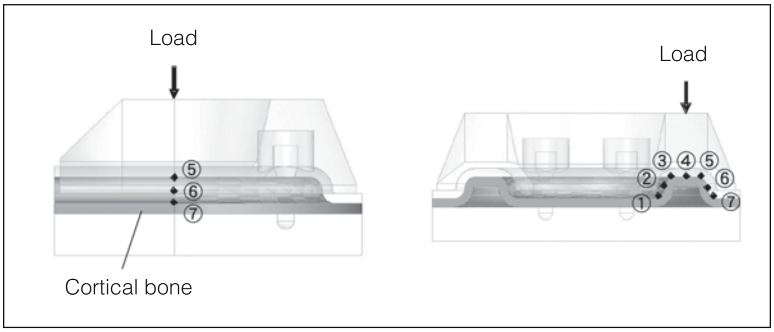

Fig 5 Stress measurement points (1 to 7) on the cortical bone. *(left)* Lateral view; *(right)* transverse view at the posterior border.

breaking material was placed around each abutment. To evaluate the influence of stress-breaking material thickness, a 1-mm-thick model (Model 1) and 3-mm thick model (Model 2) were used as the 2-implant models. Moreover, to evaluate the influence of implant number, a 1-mm thick, 4-implant model was created (Model 3). In each model, stress-breaking ability was compared by changing the elastic modulus of the stress-released materials.

Model Settings

Ten-node tetrahedral elements were used in the models. Gap elements (modulus of friction: 0.005)[12] were used at the interface between the superstructure (dentures and stress-breaking materials) and ridge (mucosa and implants). Interface nodes between each component were merged without an interface between the superstructure and ridge. The number of elements and nodes were 29,413 and 40,507 in Model 1, 38986 and 53125 in Model 2, and 44755 and 60567 in Model 3, respectively.

For efficient analysis, elements of the implants and stress-breaking material were especially fined using the 3D CAD program. Axial loads of 100 N were applied to the occlusal surface at the left first molar tooth position (Fig 4).[20,21] All nodes at the surface of the cancellous bone were restrained in all directions, as shown in Fig 4. All materials were assumed to be linearly elastic and isotropic. The material properties of the dentures, mucosa, cortical bone, cancellous bone, and implants were as reported in a previously published study (Table 1).[13] The elastic modulus of the stress-breaking material was varied from 1 to 3,000 MPa. All conditions were set using the finite element program (CosmosWorks 2005, Structure Research and Analysis).

Analyses

Maximum stress at the implant-bone interface in Models 1, 2, and 3 and stress at the cortical bone surface just under the loading point in Model 1 (Fig 5) were calculated using the finite element program (Cosmos/M Version 2.7, Structure Research and Analysis).

Results

Maximum Stress at the Implant-Bone Interface

In all models, maximum stress at the implant-bone interface in implants at the canine tooth position was generated at the peri-implant bone on the loading side (Fig 6). As the elastic modulus increased, so too did the stress; minimum and maximum stresses were

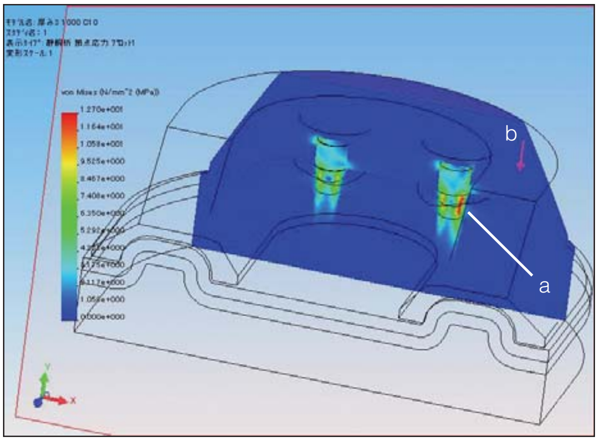

Fig 6 Stress distribution at the implant-bone interface. a: Maximum stress, b: load.

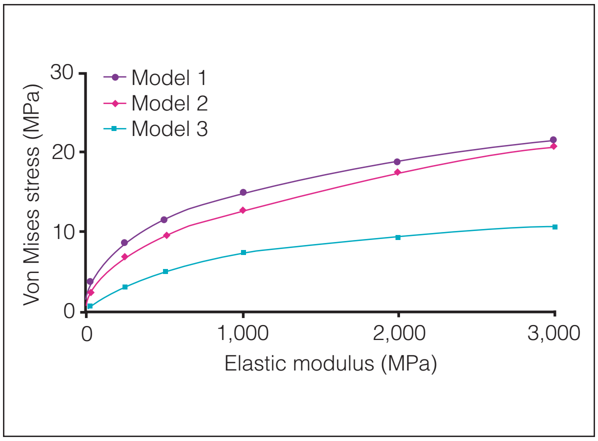

Fig 7 Maximum stress at the implant-bone interface on the loading side of the 3 models.

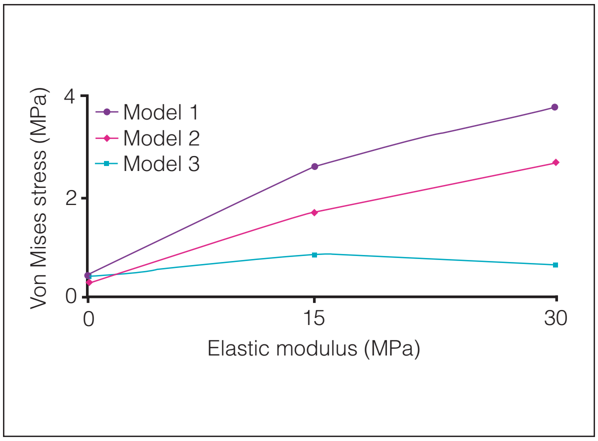

Fig 8 Maximum stress at the implant-bone interface on the loading side of the 3 models with an elastic modulus of 1 to 30 MPa.

generated with elastic moduli of 1 and 3,000 MPa, respectively (Fig 7). Moreover, with a lower elastic modulus, the rate of stress increase over the increase in the elastic modulus was even larger. The maximum rate of stress increase was observed with an elastic modulus of 1 ~ 15 MPa and the minimum with 2,000 ~ 3,000 MPa: 0.154 and 0.0025 in Model 1, 0.0986 and 0.0032 in Model 2, and 0.0318 and 0.00127 in Model 3 (Fig 8).

In the 2-implant models, stress with the 3-mm thick stress-breaking material was smaller than that with 1-mm-thick material. Moreover, compared with the 2-implant models, when the elastic modulus of the stress-breaking material was between 15 and 3,000 MPa, the stress decreased to a maximum of 82.5% at 30 MPa and a minimum of 50% at 3,000 MPa in the 4-implant model. With an elastic modulus of 1 MPa, the difference in stress between the 2- and 4-implant models was very small.

Stress on the Cortical Bone Surfaces

Regardless of the elastic modulus of the stress-breaking materials, maximum stress was generated at measurement point 7 and minimum stress at measurement point 1 (Fig 9). At points 1 and 2, maximum stress was generated with an elastic modulus of 1 MPa. As the elastic modulus increased, the stress decreased, and minimum stress was generated at 3,000 MPa. At point 1, there was an obvious difference between 1 MPa and 15 MPa. At points 3, 4, and 5, minimum stress was generated at 1 MPa, and as the elastic modulus increased so too did the stress. At point 3, maximum stress was generated at 500 MPa, while at points 4 and 5, maximum stress was generated at 250 MPa. At points 6 and 7, maximum stress was generated at 1 to 30 MPa, and as the elastic modulus increased, the stress decreased, reaching a minimum at 3,000 MPa. The sum of total stress from points 1 to 7 with each elastic modulus revealed a maximum of 692 KPa at 1 MPa and a minimum of 639 KPa at 3,000 MPa (Fig 10).

Discussion

Both the elastic modulus and thickness of the stress-breaking material affected the maximum stress at the implant-bone interface. When thicker material with a smaller elastic modulus was used, stress at the implant-bone interface was reduced. In contrast, stress on the ridge decreased with increasing elastic modulus.

It is suggested that maximum stress at the implant-bone interface is related to displacement of the denture. As the elastic modulus increases, displacement of the denture decreases. As a result, stress on the implants increases and so too does the maximum stress at the implant-bone interface. In the 2-implant models with an elastic modulus of 1 to 30 MPa, the rate of stress increase over the increase in the elastic

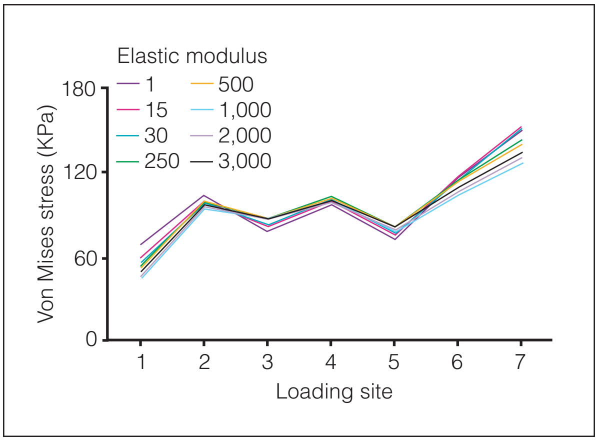

Fig 9 Stress at the cortical bone on the loading side.

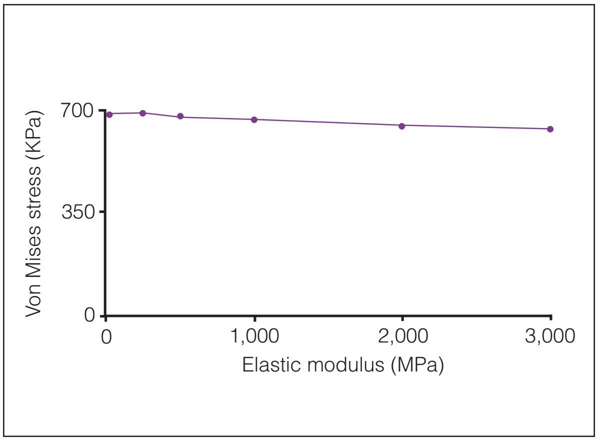

Fig 10 Total sum of stresses at the cortical bone at points 1 to 7.

modulus was obviously larger. This suggests that displacement of the denture was sensitive to this range of elastic modulus. Stress-breaking material transforms itself to reduce stress transmission from the denture to the implant; therefore, when the material is thin, the degree of transformation and thus displacement of the denture will be small. As a result, stress on the implant increases and the maximum stress at the implant-bone interface is larger than with thicker material. However, the effect of stress breaking was higher in 3-mm-thick than 1-mm-thick stress-breaking materials, while the degree of the difference was much smaller than the difference of the elastic moduli. Thus, the thickness of a stress-breaking material may have only a small influence clinically. Compared with the 2-implant models, the 4-implant model increases the number of fulcrums, resulting in a reduction in denture displacement. In the 4-implant model, the stress at each implant is small, so maximum stress at the implant-bone interface was obviously smaller than in the 2-implant models.

With maxillary implant overdentures, palatal coverage is undesirable for maintenance of oral sensation and function[2–4]; however, it has the advantages of reducing stress on the implants and improving retention and stability.[8,9,22] On the other hand, palatal coverage is not recommended with overdentures using 4 implants,[23] since it is suggested that the stress on the implants will be clinically acceptable without palatal coverage. Further, it is suggested that 2 implants are sufficient as anchors for palateless overdentures when the stress on the implants is not excessive. In the 2-implant models, the stress generated with an elastic modulus of 3,000 MPa, which is the same as the elastic modulus of acrylic resin, was similar to that at 400 MPa with 1-mm-thick material and 700 MPa with 3-mm-thick material. Therefore, to control the stress on the implants in palateless overdentures with 2 implants, it is necessary to control the elastic modulus of the stress-breaking material to below 400 MPa when the material is 1 mm thick and below 700 MPa when the material is 3 mm thick. As the elastic modulus decreased, the difference in stresses between models also decreased, and the influence of stress-breaking material thickness and number of implants was also reduced.

The stress generated at the cortical bone of the ridge suggests that when the elastic modulus of the stress-breaking material is small, displacement of the denture will be large, and thus the stress on the ridge will increase, except on the top of the ridge. This was particularly noted at an elastic modulus of 1 MPa, suggesting that the denture tended to displace on the loading side. Moreover, with a larger elastic modulus, displacement of the denture was less and the stress on the ridge decreased. From 1 to 500 MPa, it is suggested that as the elastic modulus increases, the stress on the ridge decreases, except on the top of the ridge. At 500 to 3,000 MPa, the stress on the top of the ridge decreased because of the decrease in the total sum of stresses on the ridge. Stress on the bone is a mechanical factor involved in ridge resorption.[24,25] Thus, it is suggested that to reduce the stress on the cortical bone and limit resorption of the ridge, use of a stress-breaking material with a large elastic modulus is advantageous.

Elastic moduli of 1, 15, and 30 MPa represent the elastic moduli of commercial soft, medium, and hard denture liners, respectively (Table 2).[26,27] Within the limitations of this experiment, each of these elastic moduli was effective in reducing the stress on the implants. However, at 1 MPa, the stress on the ridge increased; therefore, when using soft denture liners as the stress-breaking material, an elastic modulus of 15 to 30 MPa is recommended. If it is necessary to use a material with a higher elastic modulus, it will have to be developed, as presently no such material is available commercially. In this study, although a reduction in the elastic

Table 2 Elastic Moduli of Commercial Soft Denture Liners

Material	Manufacturer	Elastic modulus (MPa)
Evatouch Soft	Neo Dental Chemical Products	1.5
Molteno regural	Molten Medical	15.81
Molteno hard	Molten Medical	27.94

modulus of the stress-breaking material was effective in reducing the stress at the implant-bone interface, an increase was effective in reducing stress on the ridge. Because of this conflict, stress-breaking material with an elastic modulus of 15 to 400 MPa is recommended when the material is 1 mm thick and 15 to 700 MPa when it is 3 mm thick. In this study, although no retention mechanism was included to eliminate the influence of shape, the authors suggest that appropriate retention will reduce the stress on the ridge even when the elastic modulus of the stress-breaking material is low. However, more research is required to confirm this concept.

Conclusions

Within the limitations of this experiment, the following findings were revealed. The stress generated at the implant-bone interface can be controlled by the elastic modulus and thickness of the stress-breaking material. The stress is reduced with 3-mm-thick material compared to 1-mm-thick material, and as the elastic modulus decreases so too does the stress. The stress generated at the cortical bone just under the loading point is reduced as the elastic modulus increases. With maxillary implant-retained palateless overdentures with 2 implants, an elastic modulus of 15 to 400 MPa and 15 to 700 MPa is therefore recommended with 1-mm thick and 3-mm thick stress-breaking materials, respectively.

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