Nonlinear Finite Element Analysis of a Splinted Implant with Various Connectors and Occlusal Forces

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Purpose: *The aim of this study was to analyze the biomechanics in an implant/tooth–supported system under different occlusal forces with rigid and nonrigid connectors by adopting a nonlinear finite element (FE) approach.* ***Materials and Methods:*** *A model containing 1 Frialit-2 implant (placed in the second molar position) splinted to the mandibular second premolar was constructed. Nonlinear contact elements were used to simulate a realistic interface fixation between the implant body and abutment screw and the sliding keyway stress-breaker function. Stress distributions in the splinting system with rigid and nonrigid connectors were observed when vertical forces were applied to the tooth, pontic, implant abutment, or complete prosthesis in 10 simulated models.* ***Results:*** *The displacement obtained from the natural tooth increased 11 times than that of the implant, and the peak stress values within the implant system ($\sigma_{I,\ max}$) increased significantly when vertical forces acted only on the premolar of a fixed prosthesis with a rigid connector. The $\sigma_{I,\ max}$ values seen in the splinting prosthesis were not significantly different when vertical forces (50 N) were applied to the pontic, molar (implant) only, or the entire prosthesis, respectively, regardless of whether rigid or nonrigid connectors were used. Moreover, the peak stress values in the implant system and prosthesis were significantly reduced in single- or multiple-contact situations once vertical forces on the pontic were decreased.* ***Discussion:*** *The compensatory mechanism between the implant components and keyway sliding function of the implant/tooth–supported prosthesis could be realistically simulated using nonlinear contact FE analysis. The nonrigid connector (keyway device) significantly exploited its function only when the splinting system received light occlusal forces.* ***Conclusion:*** *Minimization of the occlusal loading force on the pontic area through occlusal adjustment procedures to redistribute stress within the implant system in the maximum intercuspation position for an implant/tooth–supported prosthesis is recommended.* (INT J ORAL MAXILLOFAC IMPLANTS 2003;18:331–340)

Key words: *biomechanics, bite force, dental implants, dental stress analysis, finite element analysis, mechanical stress*

Osseointegration has been accepted as the major treatment concept in implant dentistry since Brånemark presented this method (based on scientific evidence) to the North American dental community in 1982.[1–3] The use of implants has been extended from the treatment of edentulous patients to partial edentulism in clinical protocols.[4,5] However, a controversial point is whether implants should be connected to natural teeth when clinical treatment is planned.[4–6]

While splinting the implant and tooth is a rational alternative in some clinical situations, the complex biomechanical aspects of a tooth/implant–supported system are derived from the dissimilar mobility between an osseointegrated implant and a tooth. A series of physiologic and engineering problems, such as loss of osseointegration, abutment screw loosening, and prosthesis fracture, arise because of the higher bending moment caused by the cantilever effect when occlusal forces act on the system.[4,7–9] There are several methods for connecting an implant to a natural tooth that may have different capabilities for compromising the existing variations in mobility. A nonrigid connector (1 of the stress breakers) has

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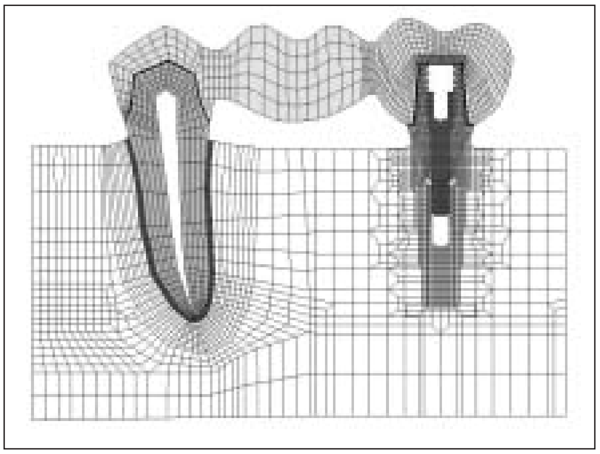


Fig 1 A plane finite element stress model of a tooth/implant–supported system was constructed for analysis in this study.

the ability to separate the splinted units, thus compensating for the different degrees of mobility between the implant and tooth. This method has been suggested for use with conventional fixed partial dentures (FPDs) by many researchers who oppose splinting the implant to a natural tooth.[5,10,11] In contrast, other studies suggest that a rigid connector is an acceptable device, because the prosthesis/implant system possesses the inherent flexibility to match dissimilar mobility characteristics.[4,12,13] Therefore, the decision to connect using a rigid or nonrigid concept remains a controversial problem when a natural tooth is splinted to an implant.

The biomechanics are recognized as the most significant factor influencing the long-term success of a tooth/implant–supported system.[14,15] Connector types, occlusal forces, and implant systems affect the detailed mechanical responses within the entire splinting system. Occlusal forces (magnitude and location) are usually the major factor that directly affects the load transfer and stress distribution. However, there has been insufficient research focusing on the relationship between the various occlusal forces and different connectors used. The major reason is that many investigations have used experimental approaches or clinical observations that could not provide enough information to determine the biomechanics of complicated tooth/implant–supported systems. Consequently, computer simulations based on the finite element (FE) method have been employed as a tool for evaluating mechanical responses, such as internal stresses or relative micromotions.

The FE method provides mechanical responses and alters the parameters in a more controllable manner; it has become a commonly used analytic tool in dental biomechanical studies.[16–20] However, simulated analytic results have been ambiguous because of unrealistic assumptions, such as interfacial fixation between the abutment screw and implant body.[17,18] Therefore, an accurate FE model with reasonable interface conditions (bonded or contact) that could simulate the inherent flexibility within the implant system and the nonrigid connector function seems necessary for computer simulations. This study was aimed at investigating the mechanical interactions in a tooth/implant–supported system under various occlusal forces with rigid and nonrigid connectors using nonlinear computer simulations.

MATERIALS AND METHODS

Finite Element Model

An adult human mandible edentulous distal to the second premolar was selected as the partially dentate model in this study. One Frialit-2 root-form implant (4.5 mm in diameter and 13 mm in length; Friadent, Mannheim, Germany) with a screw-retained MH-6 abutment (Friadent) was designed into this model at the second molar position. This implant system (ie, an implant, abutment, and abutment screw) was embedded into a prepared machine cube with epoxy resin. The cube was sliced to expose the implant-resin section perpendicular to the occlusal surface using a slicer. The detailed dimensions of the implant, abutment, and abutment screw were measured through image scanning using a flatbed scanner. A 2-dimensional plane stress model, symmetric about the mid-pontic and including the implant, natural tooth, periodontal ligament, and fixed partial denture prosthesis, was constructed in an FE package (ANSYS version 5.6; Swanson Analysis, Houston, PA) to perform the computer simulation. This model consisted of 3,298 quadrilateral elements and 3,659 nodes (Fig 1). To simulate the interfacial fixation of different connectors, a mapping approach was adopted to design the mesh pattern according to the characteristics of the section geometry for the keyway stress breaker (Fig 2). The exterior nodes of the alveolar bone in the FE models were fixed in all directions as the boundary conditions. The material properties of the dental tissues, alveolar bone, prosthesis, and implant materials were adopted from the literature (Table 1).

Interfacial Fixations Within the Implant System

A vertical force (50 N) was applied first to the natural tooth cusps as loading condition type 1 (Table 2) to evaluate the mechanical responses within the

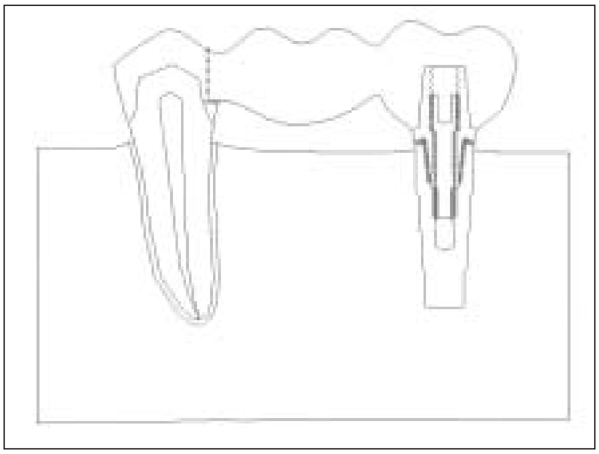


Fig 2 Nonlinear contact elements *(asterisks)* were used to simulate the realistic interface fixation with the implant system and the sliding keyway stress-breaker function.

Table 1 Material Properties Assigned to Dental Tissues, Alveolar Bone, Prosthesis, and Implant Material

Material	Young's modulus (MPa)	Possion's ratio	References
Dentin	18,600	0.31	Middleton et al[21]
Periodontal ligament	170	0.45	Weinstein et al[22]
Alveolar bone	3,430	0.30	Carter and Hayes[23]
Gold alloy (prosthesis)	90,000	0.30	Benzing et al[24] and Moffa et al[25]
Titanium (implant system)	110,000	0.35	Benzing et al[24] and Van Rossen et al[26]

tooth/implant–supported system for different implant interfacial fixations under the 1-piece prosthesis (rigid connector) design. Two interfacial conditions, bonded and contact fixations, were modeled to simulate the adaptation between various components within the implant system (abutment/implant, abutment screw/abutment, and abutment screw/ implant) (Fig 2). The bonded fixation meant that relative micromotions were not allowable, and displacement was continuous between different materials. In contrast to the bonded fixation, contact fixation is one of the nonlinear structural analyses provided in ANSYS. The contact element (defined as node to surface) allowed nodes to slip in the tangential direction with no penetration between different materials. The magnitude of the relative micromotions and compressive stress transfer could be obtained directly from the contact simulation problems. There were 1,084 additional contact elements assigned to model the interfacial fixation between the various components within the implant.

Connector Designs and Occlusal Forces

Under contact fixation between different components within the implant system, 5 loading types with 2 connector designs (rigid and nonrigid) were considered as the calculated modes to understand the stress distributions in the alveolar bone, the prosthesis, and the implant system. Nonlinear contact elements (n = 13) were also employed to simulate the sliding keyway stress breaker function (Fig 2). There were 11 simulated models in this study. Table 2 lists the detailed loading positions, connecting types, and interfacial fixations of these FE models.

RESULTS

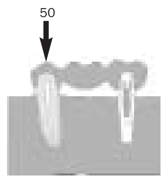
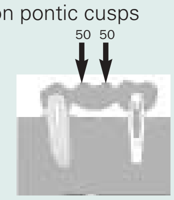
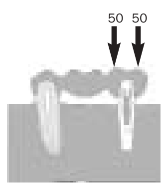
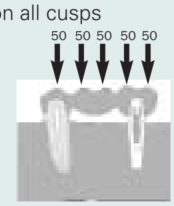
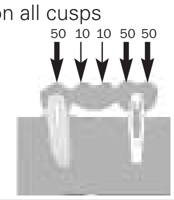
Interface Effect Within the Implant System

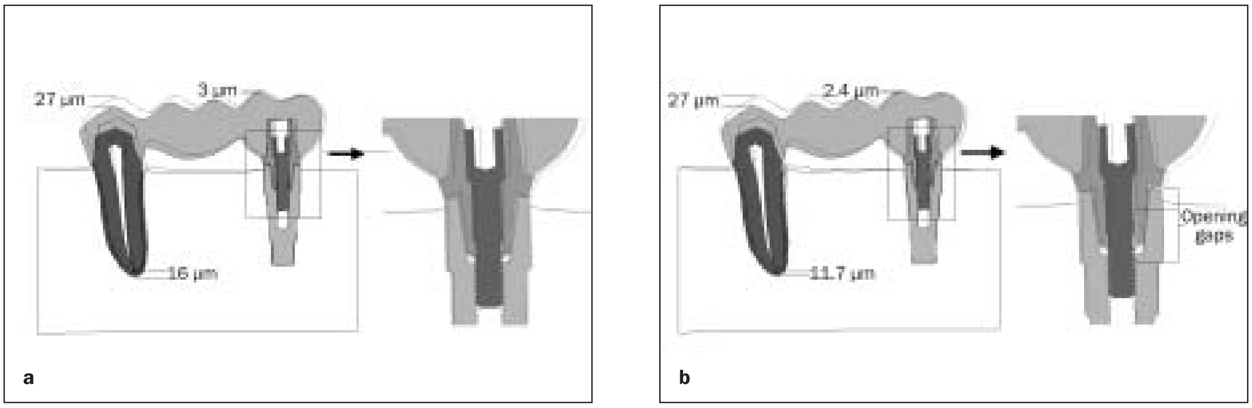
When occlusal force was applied to the premolar cusp of the fixed prosthesis (loading type 1) with the rigid connector, the magnitude of the tooth cusp displacements was similar for bonded and contact fixation models (Figs 3a and 3b). The displacement obtained from the natural tooth increased 9 to 11 times than that of the values in the implant for the bonded and contact fixations as a result of the dissimilar mobility between the implant and tooth. Smaller molar cusp displacements (implant side) and opening gaps (magnified 100 times) were also observed between different components within the implant system when the contact condition was modeled in the simulation (Fig 3b). Owing to the relative micromotions among the different components and the discontinued displacements, the distributed stresses within the implant system with contact conditions were lower than stresses for bonded case patterns (Figs 4a and 4b). As a result of this analysis indicating that the contact mode displayed more realistic interface fixation, the following simulated results were presented under contact fixation conditions between various components within the implant system.

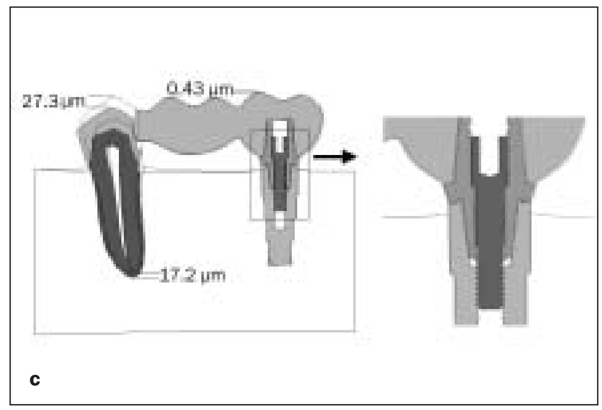
Different Connectors with Various Occlusal Forces

When occlusal force was applied to the premolar cusp (loading type 1), the tooth displacement values were similar for the prosthesis with rigid (27 μm) and nonrigid (27.3 μm) connectors. For the implant side, the displacement obtained from the prosthesis

Table 2 Detailed Loading Positions, Connecting Types, and Interfacial Fixations of Simulated FE Model Types in this Study

Loading type	Connecting type	Implant component interface fixation
1 One vertical load (50 N) on premolar cusp 	Nonrigid (with keyway)	Contact
	Rigid (without keyway)	Contact
	Rigid (without keyway)	Bonded
2 Two vertical loads (50 N) on pontic cusps 	Nonrigid (with keyway)	Contact
	Rigid (without keyway)	Contact
3 Two vertical loads (50 N) on molar cusps 	Nonrigid (with keyway)	Contact
	Rigid (without keyway)	Contact
4 Five vertical loads (50 N) on all cusps 	Nonrigid (with keyway)	Contact
	Rigid (without keyway)	Contact
5 Five vertical loads (10 N on pontic and 50 N on teeth) on all cusps 	Nonrigid (with keyway)	Contact
	Rigid (without keyway)	Contact





Figs 3a to 3c Under an occlusal force acting on the premolar cusp (loading type 1), displacements (magnified 100 times) were calculated for 3 different models: *(a)* rigid connector (without keyway) and bond fixation within the implant system for different components; *(b)* rigid connector (without keyway) and contact fixation within the implant system; *(c)* nonrigid connector (keyway designs) and contact fixation within the implant system.

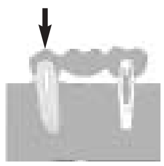
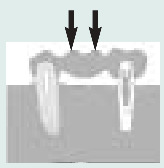
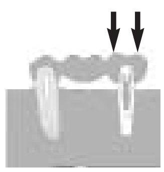
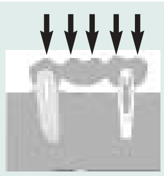
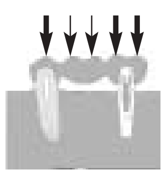
with a nonrigid connector was only 0.43 µm while the premolar was under loading. This displacement value is only one fifth that of the prosthesis with a rigid connector (2.4 µm/0.43 µm) (Figs 3b and 3c). The peak von Mises stress value within the implant ($\sigma_{I,\ max}$) decreased significantly, from 77.56 MPa to 3.2 MPa, for rigid and nonrigid connectors (Table 3, Figs 4b and 4c). This was because the keyway device broke the stress transfer from the natural tooth to the implant side. However, there were no significant differences in the peak von Mises stresses for alveolar bone ($\sigma_{B,\ max}$) and the prosthesis ($\sigma_{P,\ max}$) regardless of whether rigid or nonrigid connectors were used (Table 3).

When occlusal forces were applied to the pontic, molar cusps, or all cusps of the splinting system (loading types 2, 3, and 4), increased stress values were found within the implant system for the prosthesis with a nonrigid connector. This was because the stress could not be transferred to the natural tooth, so the implant system received a higher axial load (loading types 3 and 4) and bending moment (loading types 2 and 4) (Table 3, Figs 5a to 5f). No significant differences in the peak stresses were found for the alveolar bone and prosthesis with different types of connections. When the occlusal forces acting on the pontic were adjusted to be smaller than the forces on the other areas (the proportion was 1:5, loading type 5), the values for $\sigma_{I,\ max}$ and $\sigma_{P,\ max}$ decreased significantly. The maximum peak stress for loading type 5 was only half that of loading type 4 for rigid or nonrigid connectors because of the lower bending moment effect in the implant system (Table 3, and Figs 5e to 5h).

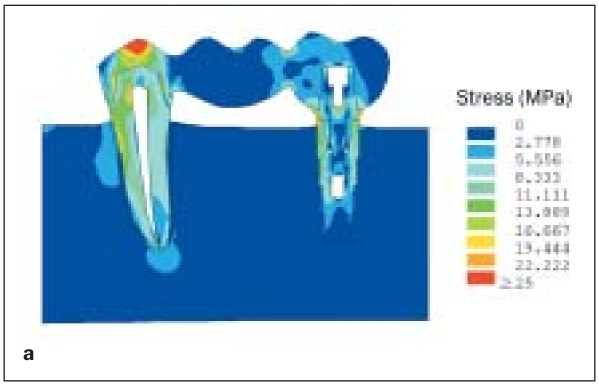
DISCUSSION

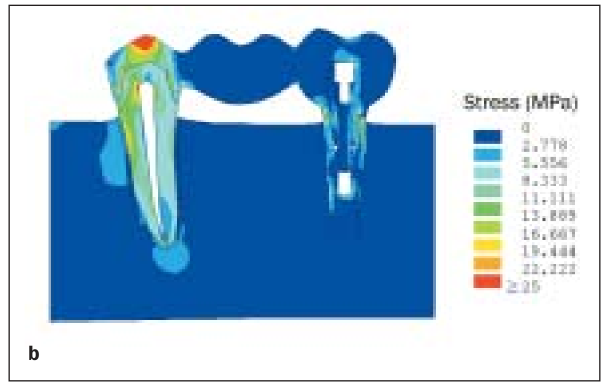
Although implant and tooth splinting is considered a rational alternative in some clinical situations, the long-term physiologic and engineering problems induced by the difference in mobility between the tooth and implant have been controversial for clinical treatment protocols. Despite many devices such as nonrigid connectors and flexible implant systems proposed to balance the differences in mobility, the

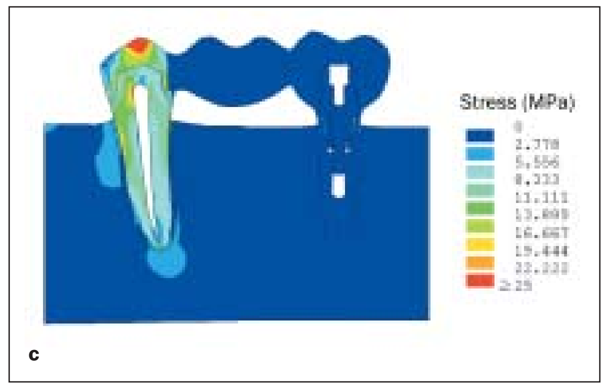
Table 3 Contact-simulated Results of Peak Von Mises Stresses of Implant System, Alveolar Bone, and Prosthesis for 2 Connectors with 5 Loading Types

Loading type	Connecting type	Peak stress in implant system (MPa)	Peak stress in alveolar bone (MPa)	Peak stress in prosthesis (MPa)
1 	Nonrigid (with keyway)	3.2	5.62	53.67
	Rigid (without keyway)	77.56	5.32	53.63
2 	Nonrigid (with keyway)	135.89	6.12	306.45
	Rigid (without keyway)	122.60	5.80	94.00
3 	Nonrigid (with keyway)	95.26	11.36	99.52
	Rigid (without keyway)	93.67	11.43	99.55
4 	Nonrigid (with keyway)	207.4	15.27	297.53
	Rigid (without keyway)	195.23	15.34	148.68
5 	Nonrigid (with keyway)	103.25	10.81	99.92
	Rigid (without keyway)	101.50	10.82	99.89

feasibility and efficacy of these methods require further evaluation because the basic biomechanical aspects are still unclear. In 1983, Skalak proposed that biomechanics were among the factors influencing the long-term success of tooth/implant–supported systems.[14] However, the biomechanical aspects are difficult to evaluate using clinical observation/experimental approaches with limited information and sample variations. Therefore, FE analysis has generally been accepted as a complementary tool for understanding the detailed mechanical responses for many biologic investigations. The accuracy of an FE analysis is dependent upon the numeric convergence and correctness of the assumptions imposed on the models simulating real physical conditions, eg, the boundary conditions and interfacial conditions. Consequently, nonlinear contact analysis was needed to mimic a flexible implant system and keyway sliding function and provide additional information for clinical consideration.

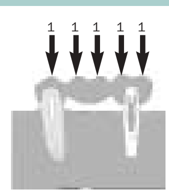
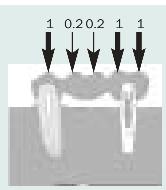






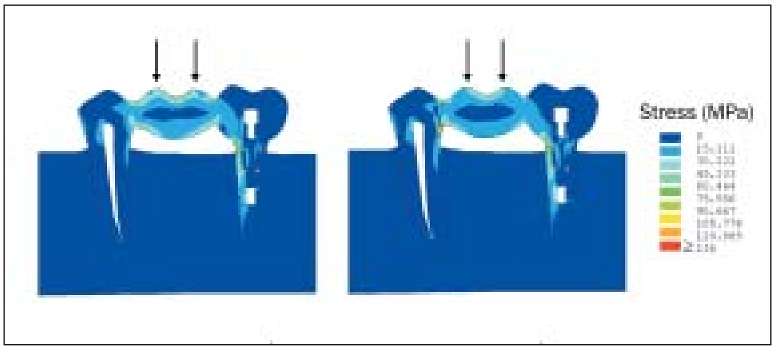
Figs 4a to 4c Von Mises stress distributions for *(a)* the rigid connector used with bond fixation within the implant system, *(b)* the rigid connector used with contact fixation with the implant system, and *(c)* the nonrigid connector used with contact fixation within the implant system under loading type 1 condition.

Table 4 Peak Von Mises Stresses of Implant System, Alveolar Bone, and Prosthesis for 2 Connectors with Loading Types 9 and 10

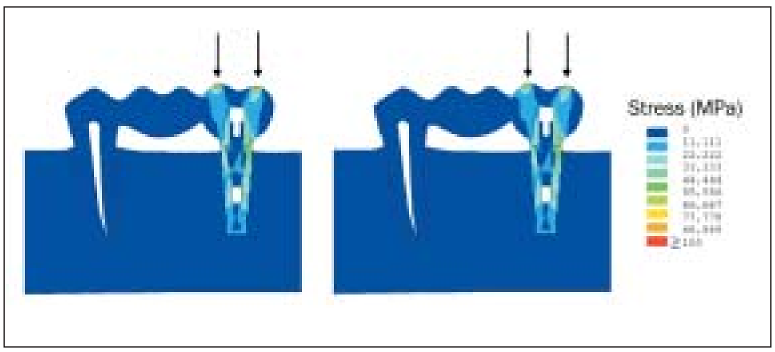
Loading type (N)		Connecting type	Peak stress in implant system (MPa)	Peak stress in alveolar bone (MPa)	Peak stress in prosthesis (MPa)
9		Nonrigid (with keyway)	37.05	2.02	18.91
		Rigid (without keyway)	53.93	2.84	24.75
10		Nonrigid (with keyway)	4.73	0.43	3.67
		Rigid (without keyway)	30.48	1.67	14.55

The results obtained from contact fixation analysis indicated that the compensatory mechanism within the implant system for dissimilar mobility was simulated more realistically for the relative micromotions that occur between various components. When compared with the interfacial bond conditions, the stress values in the abutment, abutment screw, and implant also decreased, because the opening gaps prevented the transfer of stress from the prosthesis to the implant system (Figs 4a and 4b). This result is consistent with a previous experimental study that used the bench test approach to

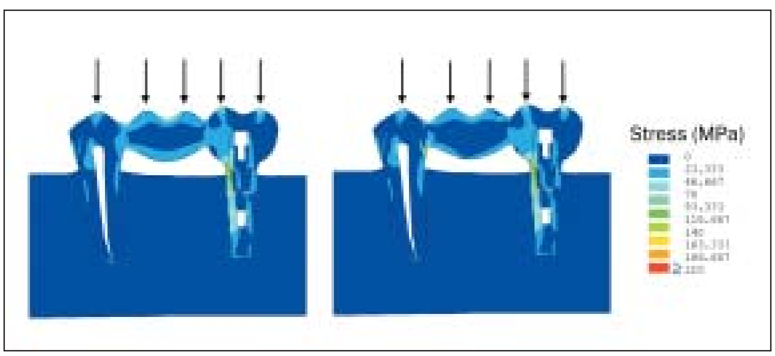
Figs 5a to 5h *Von Mises stress distributions for rigid and nonrigid connector types of tooth/implant–supported system with loading types 2 to 5.*



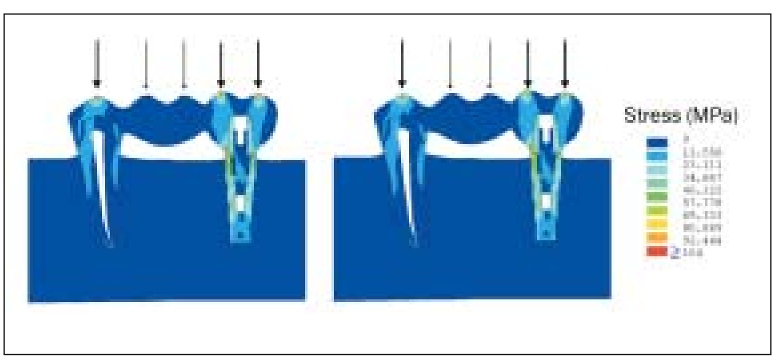
Figs 5a and 5b Von Mises stress distributions in the simulated tooth/implant–supported system under loading type 2 with *(left)* rigid and *(right)* nonrigid connections.



Figs 5c and 5d Von Mises stress distributions in the simulated tooth/implant–supported system under loading type 3 with *(left)* rigid and *(right)* nonrigid connections.



Figs 5e and 5f Von Mises stress distributions in the simulated tooth/implant–supported system under loading type 4 with *(left)* rigid and *(right)* nonrigid connections.



Figs 5g and 5h Von Mises stress distributions in the simulated tooth/implant–supported system under loading type 5 with *(left)* rigid and *(right)* nonrigid connections.

evaluate the flexibility of a Brånemark System implant.[7]

For prostheses with different connectors loading with various occlusal forces, the value for $\sigma_{I,\ max}$ with a rigid connector showed stress values that were 24 times greater than those seen for the nonrigid connector when occlusal force acted on the natural tooth (loading type 1) (Table 3). This was the result of stress transfer from the loading site to the implant system when a rigid connection was used. Lower stress values were also found in implants with rigid connectors and loading types 2, 3, 4, and 5, because the natural tooth shared the stress via transfer through the rigid connector

(Table 3). These analyses suggested that nonrigid connectors, ie, keyway devices, play the role of stress breaker within the tooth/implant–supported system. However, different connector types are not the only factor influencing stress transfer in the whole system. A higher bending moment occurring at the implant side caused by particular occlusal forces (loading types 2 and 4) is a significant issue that affects stress redistribution. When occlusal forces were applied to the molar cusps only (loading type 3), the implant side received most of the axial forces, and the difference in peak stress values was only 1.67% ([95.26 – 93.67]/95.26) (Table 3) between rigid and nonrigid connections. In contrast, 9.7% ([135.89 – 122.6]/135.89) and 5.8% ([207.4 – 195.23]/207.4) (Table 3) differences between the keyway device and 1-piece prosthesis were found for loading types 2 and 4, respectively, because they generated higher bending moments on the implant system. The stress values near the bottom contact areas of the keyway device for load types 2 and 4 increased dramatically, to around 300 MPa (Table 3). Therefore, the failure risk for the prosthesis might rise profoundly after long-term dynamic loads. From these simulated results, it was apparent that both the occlusal contact force and contact location affected the stress distribution in the implant/tooth–supported prosthesis with different connector designs. To reduce the cantilever effect, the occlusal forces acting on the pontic were reduced (loading type 5) to revaluate the stress distribution. The results indicated that the peak stress values in the implant decreased significantly, and a 1.69% ([103.25 – 101.5]/103.25) variation (Table 3) was obtained for different connector types. This finding implied that further understanding the role of occlusal adjustment can affect the long-term success of a tooth/implant–supported prosthesis.

The tooth mobility capability and biomechanical aspects of the tooth/implant–supported system were influenced significantly by the viscoelastic property of the periodontal ligament (PDL). Linear elastic (homogeneous and isotropic) PDL characterization was only involved in this study as the result of numeric convergence considerations and a larger variation for the PDL physical properties in the literature. However, the mechanical behavior of the PDL also changes nonlinearly, depending on the magnitude and duration of the load applied when the PDL is assumed to be linearly elastic.[27,28] Yoshida and coworkers[27] and Provatidis[28] pointed out that Young's modulus of elasticity for the PDL increased almost exponentially with the load increment. This value was found to be approximately 0.68 MPa under a load of 1 N (ie, orthodontic treatment).[27,28] Young's modulus of the PDL (170 MPa) assumed in this study was considered a static, heavy occlusal loading situation. To understand the cantilever effect of a tooth/implant– supported system with a nonlinear PDL property, 0.68 MPa of the Young's modulus and a reduction in the previous loading conditions (loading types 1 to 5) by 50 times (1 N and 0.2 N) were employed to perform another 10 analyses (loading types 6 to 10 with rigid and nonrigid connections). The results indicated that the stress distribution tendencies were similar when occlusal forces were applied on the premolar, pontic, and implant sides (loading types 6 to 8). When occlusal forces were acting on the entire system (loading type 9), the value for $\sigma_{I,\ max}$ with a 1-piece prosthesis was higher than with the keyway device, because the modulus of elasticity of the PDL was nonlinearly reduced (Table 4). A lower elastic modulus induces dramatic tooth intrusion and magnifies the cantilever effect for the 1-piece prosthesis system. After the occlusal forces on the pontic were reduced (loading type 10), the $\sigma_{I,\ max}$ values also decreased significantly. This result also implied that occlusal adjustment might be an important issue influencing the mechanical aspects when a tooth/implant–supported system receives smaller occlusal forces.

Nonlinear contact elements were applied in this study to simulate interface fixation within the implant system and keyway device sliding function. However, based on the limited 2-dimensional simulation data, some of the realistic physiologic and engineering factors, such as buccolingual occlusal forces, lateral keyway sliding, and hexagon fixative function within the implant, cannot be simulated in this situation. The results of all simulations were independent of the buccolingual dimension and interpreted qualitatively. Because a 2-dimensional stress model cannot simulate alveolar bone connections around the tooth and implant, only 1 material was assumed for the alveolar bone to perform the analyses in this study. The bone stress values obtained from all simulations were the first approximation. Therefore, overloading-induced loss of osseointegration was not discussed in this study. To clarify the biomechanical aspects of a tooth/implant splinting system, 3-dimensional FE analysis and clinical trials are needed.

CONCLUSIONS

Based on a 2-dimensional nonlinear FE analysis of an implant/tooth–supported system with rigid and nonrigid connectors, the following conclusions were made:

1. Nonlinear contact analysis could be employed to more realistically simulate the compensative mechanism within the implant system and keyway sliding function.
2. Both the occlusal contact force and contact location affected the stress distribution in a splinting system with different connector designs under a static, heavy occlusal loading situation.
3. The stress-breaking function of the keyway device becomes obvious only when the splinting system receives smaller occlusal forces.
4. Occlusal adjustment procedures can reduce the cantilever effect and redistribute stress within the implant system.

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