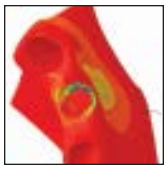
Finite Element Analysis Relative to the Crestal Position of a 3.0-mm-Diameter Implant



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It has been shown that implant designs and different vertical positions have an influence on crestal bone. The purpose of this study was to use finite element (FE) analysis to biomechanically investigate the influence of the stress/strain distribution in a maxillary anterior 3.0-mm-diameter implant in relation to its apicocoronal level after oblique loading. Two different FE models, depending on implant position relative to bone crest, were applied. It can be concluded that placing the implant-abutment interface supracrestally provides decreased levels of stress and strain in the surrounding bone. However, placing the implant 0.5 mm supracrestally is also acceptable according to this analysis.
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Recent studies have shown that implants used for replacing missing teeth have a high success rate with marginal bone loss. The bone loss associated with a two-piece implant is approximately 2 mm apical to the implant-abutment interface.[1–3]

Several reasons may explain the bone loss observed 1-year postloading.[4,5] One important factor was the apicocoronal position of the implant-abutment interface relative to the bone crest.[6,7]

The location of the implant-abutment interface is also important for the final esthetic reconstruction. Usually the implant is placed apically from the crestal bone. This can produce more bone loss that can compromise the final esthetic result.[8]

Occlusal overload has been demonstrated in some animal studies and can contribute to implant failure. It has been suggested that excessive dynamic loads might cause craterlike bone defects around osseointegrated implants.[9,10] This situation can be explained by the "mechanostat theory" developed by Frost in 1994.[11] Based on previous Frost

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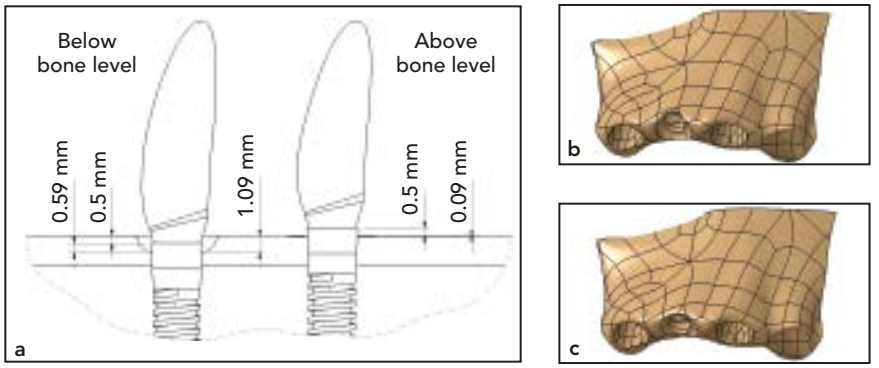

Fig 1 (a) *Geometric design of implant-abutment interfaces. Apicocoronal implant position. Crestal bone loss induced by implant position after 1 year of loading.* (b) *Below bone level model.* (c) *Above bone level model.*

Table 1 Implant and abutment designs

	Size	Part no.	Length (mm)	Material
Implant	3.0 mm	TP312L	12	Commercially pure titanium
Abutment	3.0 mm, 0 deg	TP3SEA	–	Titanium alloy (Ti6Al-4V ELI)

studies, five types of strain levels interrelated with different load levels in bone were described: *(1)* disuse, bone resorption; *(2)* physiologic load, bone homeostasis; *(3)* mild overload, bone mass increases; *(4)* pathologic overload, irreversible bone damage; and *(5)* fracture.[11]

Finite element analysis (FEA) is an engineering method that allows scientists to evaluate the biomechanics of bone around implants.[12,13] Through the use of an FEA model, it is possible to simulate different clinical situations and evaluate the best option from a biomechanical point of view. It has been reported that an inadequate bone stress distribution may inhibit bone ingrowth, induce the formation of a fibrotic layer, provoke bone loss in the locations where the force is, and, finally, increase the risk for implant loss.[14]

The purpose of this study was to biomechanically analyze, for the first time in the literature, the influence of the stress/strain distribution in bone around an anterior maxillary 3.0-mm-diameter implant in relation to its apicocoronal position after an oblique loading.

Method and materials

A model of a maxillary segment in the incisal region featuring an implant and its suprastructure was constructed using a computer-aided design program (Catia V5 R20, Dassault Systèmes). The cortical bone was assumed to have a mean thickness of 2.0 mm.[15] The human maxilla model was reconstructed using a computed tomography imaging technique.

The geometry of the implant and abutment designs was obtained from mechanical drawings provided by the manufacturer (Bio-Horizons) (Table 1). A 3.0 × 12-mm Laser-Lok 3.0 cylindrical implant design without laser microgrooving surface treatment, made of commercially pure titanium, was used for this study. The implant was placed in the maxillary left lateral incisor area. The overall dimensions of the crowns were 6.7 mm in mesiodistal length and 4.7 mm in buccolingual width. The crown height was 11.5 and 11 mm for the subcrestal and supracrestal implant positions, respectively. The crown was attached to a 3-mm-high, straight esthetic implant abutment.

The models were classified into two groups, depending on the vertical position of the platform: above bone level or below bone level.[9] When the platform was located 0.5 mm above the bone level, it was considered to be above bone level. If it was located 0.5 mm below bone level, it was categorized as below bone level (Fig 1).

The relationship between bone changes and position of implant placement was taken into consideration. The distance between the microgap and the crestal bone 1 year after loading can produce a crestal bone loss of 0.59 mm.[16] However, when a regular two-stage implant was placed, bone loss was increased to a range of 1.5 to 2.5 mm.[17] Both FEA models were designed with a saucerization defect located around the neck of the implant, simulating the crestal bone loss, to

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investigate the effect of bone loss on mechanical responses. The below bone level model showed the largest bone changes, whereas the above bone level model showed the smallest (see Fig 1).

All materials used in this study were considered to be isotropic, homogeneous, and linearly elastic. The mechanical properties of different components used in this study were adopted from the literature, as shown in Table 2.[18–22] An elastic modulus of 102,000 MPa and Poisson ratio of 0.33 were used for the commercially pure titanium.[18–20] The elastic modulus and Poisson ratio for the cortical bone were 13,000 MPa and 0.30, respectively.[22] For cancellous bone they were 1,370 MPa and 0.30, respectively.[21] The elastic modulus and Poisson ratio for the gold alloy were 90,000 MPa and 0.30, respectively.[18–23] Since the elastic modulus and Poisson ratio of commercially pure titanium are almost the same as those of titanium alloy,[23] a model with unified abutment and implant areas (one-piece type) was designed for the analysis.

The interface between cortical bone and cancellous bone and bone-implant interface was assumed to be perfectly bonded, simulating 100% osseointegration. The crown and implant-abutment were assumed to be connected as a single unit, without any loosening.

The model was constrained in all directions at the nodes on the distal end, mesial end, and upper surface of the bone segment. A mastication force[15] was simulated as an oblique force of 100 N in a 30-degree[24] direction (relative to the long axis of the implant from palatal to labial direction) (Fig 2). Loading was applied at 2-mm overbite apically on a node at the fossae of the crown to simulate clinical conditions. Oblique occlusal forces represented a more realistic situation of the oral environment and for a given force will cause the highest localized stress in cortical bone.[13]

Analysis and postprocessing were performed for each model using an FEA method program (ABAQUS, Dassault Systèmes Simulia).

The principal stress and strain distributions around the implant-bone interface were displayed along the labialpalatal section.

Fig 2 *Direction and location of a 100-N oblique load force. Vertical component of the load equals 86.6 N, and horizontal component of the load equals 50.0 N.*

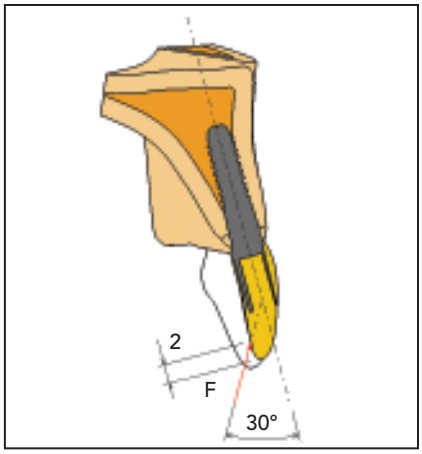

Table 2 Physical properties of different components used

Material	Elastic modulus (MPa)	Poisson ratio	Reference
Cortical bone	13,000	0.30	Carter and Spengler[22]
Cancellous bone	1,370	0.30	Sevimay et al[21]
Commercially pure titanium (implant, abutment)	102,000	0.33	Benzing et al[18] Van Rossen et al[20]
Gold alloy (crown)	90,000	0.30	Benzing et al[18] Moffa et al[19]

Results

For the cortical bone in both models, the tensile stresses were concentrated on the palatal side of the implant neck and the cortical bone. Peak values of tensile stresses were 58.35 MPa for the below bone level model and 42.55 MPa for the above bone level model (Fig 3). Taking into consideration that 100% force equals the fracture of the bone, these forces were 67.10% and 48.93%, respectively.

Compressive stresses were concentrated on the labial side of the implant neck and cortical bone. Peak values were –35.14 MPa for the below bone level model and –56.16 MPa for the above bone level model (Fig 4). This equates

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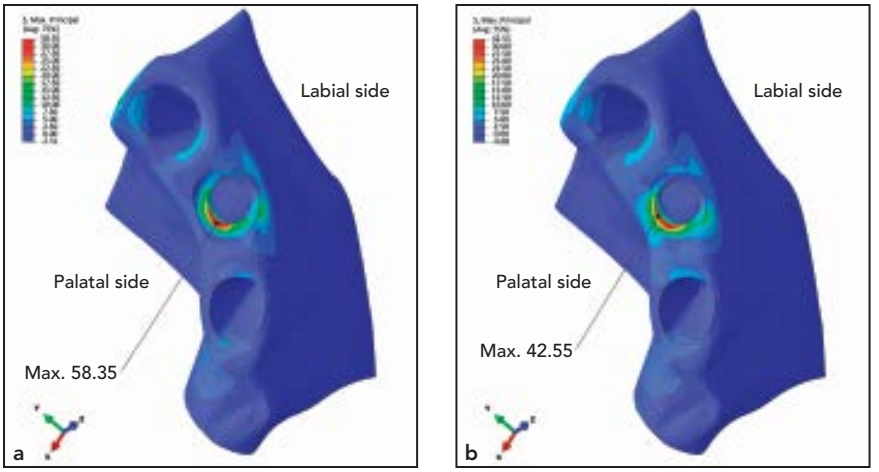

Fig 3 *Maximum principal (tensile) stress (MPa) in compact bone after an oblique force of 100 N is applied.* (a) *Below bone level model.* (b) *Above bone level model. Red and orange areas indicate high intensity stresses. Area of maximum tensile stress is indicated. Note that the implant is excluded from the illustrations for clarity.*

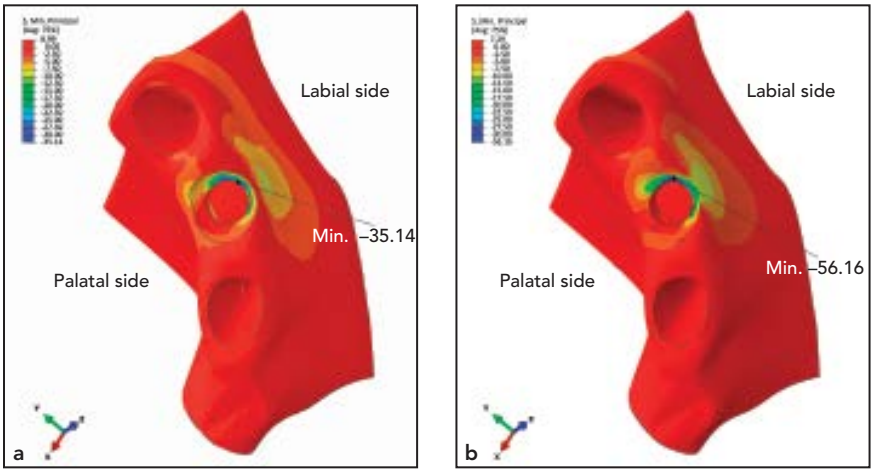

Fig 4 *Minimum principle (compressive) stress (MPa) in compact bone after an oblique force of 100 N is applied.* (a) *Below bone level model.* (b) *Above bone level model. Blue and green areas indicate high intensity stresses. Area of minimum compressive stress is indicated.*

to a force of 19.52% and 31.20%, respectively.

In the below bone level model, tension suggested a stress increase of 37.13% while the above bone level model had a compression of 59.82%.

For cancellous bone, the peak values of tensile stresses were 2.32 MPa for the below bone level model and 2.50 MPa for the above bone level model (Fig 5). This equates to 46.40% and 50.0%, respectively.

Compressive peak values were –3.72 MPa for the below bone level model and –2.70 MPa for the above bone level model (Fig 6). This equates to a force 74.60% and 53.80%, respectively.

In the below bone level model, tension suggested a stress increase of 7.20%, and in the above bone level model, compression was 38.66%. The minimum principal stress value for cancellous bone was 74.0% for below bone level and 54.0% for above bone level.

The maximum equivalent strains were located at the top of crestal cortical bone around the implant neck buccally in both models (Fig 7). The below bone level model showed a peak value of 3,045.13 με which was 2.34% higher than the above bone level model (2,975.56 με). Both the above and below bone level models were below the level of pathologic strain. The maximum equivalent strain of the below bone level model was 76.13%, and for the above bone level model, the pathologic strain level was 74.39%.

Stress distributions at the implant were generated at the abutment-implant interface and the first thread of the implant. The compression side of the implant showed the highest stress magnitudes for both the above (125.3 MPa) and below (128.0 MPa) bone level models. The apical position model had higher stress at the implant-abutment interface; however, comparable stress distribution and intensity were seen on the implant-abutment interface for both models (2.14%). The maximum stress values for the implant body were 19.27% and 19.69%, respectively, and these forces were lower than the titanium alloy (650 MPa) (Fig 8).

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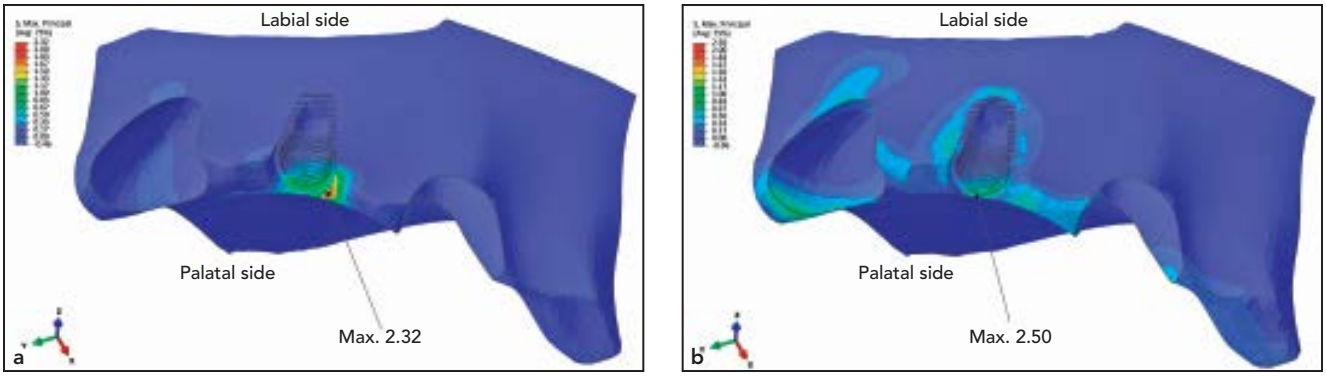

Fig 5 *Maximum principal (tensile) stress (MPa) in cancellous bone after an oblique force of 100 N is applied.* (a) *Below bone level model.* (b) *Above bone level model. Area of maximum tensile stress is indicated.*

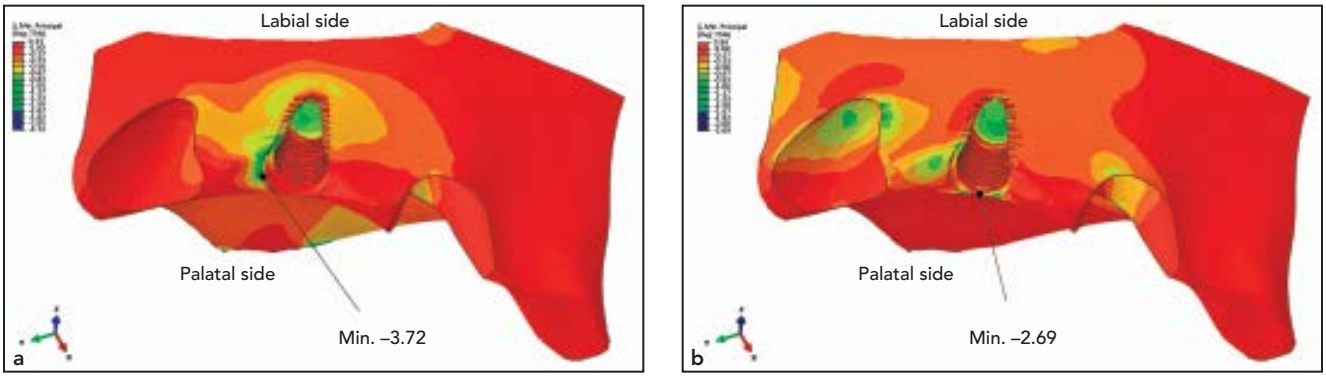

Fig 6 *Minimum principal (compressive) (MPa) in cancellous bone after an oblique force of 100 N is applied.* (a) *Below bone level model.* (b) *Above bone level model. Area of minimum compressive stress is indicated.*

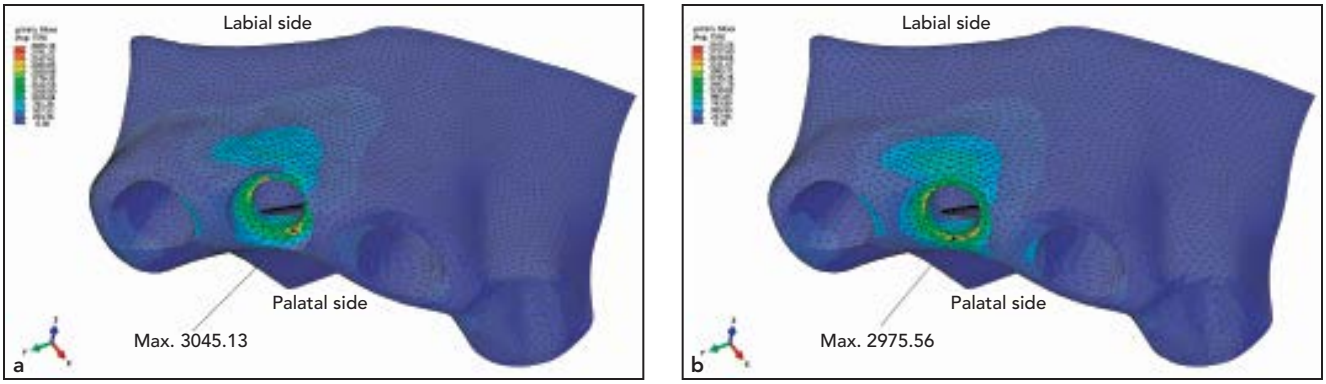

Fig 7 *Equivalent microstrains in compact bone after an oblique force of 100 N is applied.* (a) *Below bone level model.* (b) *Above bone level model.*

Discussion

The results demonstrate that the apicocoronal position of the implant-abutment interface significantly affects the magnitude of the tensile and compressive peak stresses in cortical bone. This FEA study indicated that peak values of periimplant compressive stress in compact bone were lower in the above bone level model (–56.16 MPa) compared with the below bone level model (–35.14 MPa) and occurred mainly at the labial surface of compact bone adjacent to the first thread of the implant, near the junction of compact and

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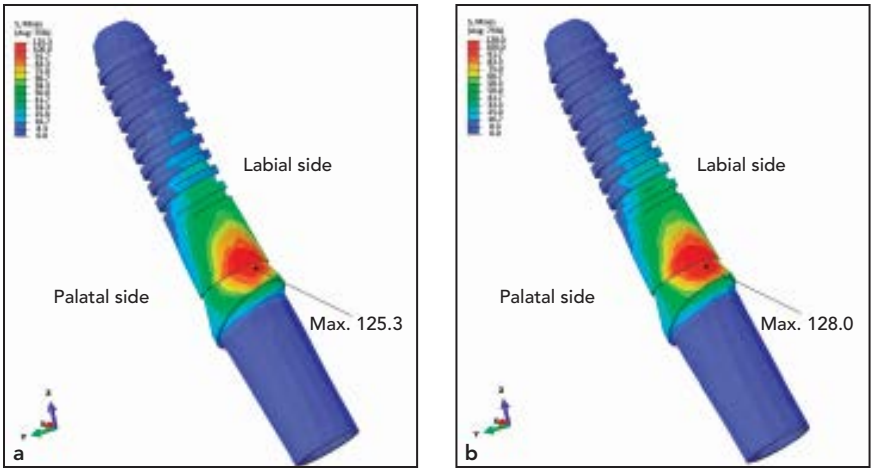

Fig 8 *von Mises stress (MPa) distribution on the implant-abutment interface.* (a) *Above bone level model.* (b) *Below bone level model. Deformation is magnified ×50 for visualization.*

cancellous bone in both models. It has been shown that compression may compromise the periosteal blood supply[25] and lead to necrosis and that high compressive stresses may increase the risk of bone loss,[26] while extensive tensile stress has also been reported to cause bone resorption.[26,27] The apparent stress concentration was reduced in compact bone, and the stress distribution in cancellous bone was shifted along the entire surface of the screw thread in both models.

According to Frost's theory, the compressive and tensile stress levels applied to the cortical bone were confirmed to be safe for clinical use. The FEA results imply that under the same magnitude of oblique loading, a benefit of decreasing the surrounding bone strain occurs when placing the implant in a supracrestal position (3,045.13 με) compared to a subcrestal position (2,975.56 με). Both models shown in Fig 8 are within the physiologic overload zone that covers the range between 2,000 and 4,000 με and is suggested to result in an increase of bone mass.

In this respect, the equivalent strain offers an effective means for measuring the overall tissue deformation gradient of the biologic response to mastication in the mandibular bone. The literature has shown that strain is perhaps one of the most relevant measures for predicting bone remodeling, based upon Frost's remodeling theory.[28]

As von Mises criteria confirms, excess strength produces implant failure, and implant yield failure was predicted in both the above and below bone level models under a static oblique load of 100 N. The maximum implant-abutment von Mises stress of 128.0 MPa for the below bone level model was well below the 650 MPa tensile yield limit of commercially pure titanium. This may be attributed to the bending moment being greater in the below bone level model with a subcrestal 0.5-mm implant position because the entire length of the bone is not in contact with the implant, which means a longer resistance arm. These results suggest that the Laser-Lok 3.0 implant-abutment design should be able to sufficiently withstand a 100-N oblique force. However, dynamic effects may add up to about 10% to 20% or more loading, which must be taken into account to avoid fracture or fatigue failure of the implant.

A limitation of the present study was the use of a static occlusal force in the analysis. Although an oblique static load has been suggested to represent a realistic occlusal load,[29] chewing movement, especially with dynamic loading simulations, needs to be considered in future investigations.

Conclusions

Based on the results of the analyses and considering the limitations of this study, the following conclusions can be drawn: *(1)* the position of the implant-abutment interface has an important role on the stress/strain distribution in peri-implant bone, *(2)* the FE analysis shows that the equivalent strains in both scenarios avoid overloading the alveolar bone surrounding implants, according to Frost's mechanostat theory; therefore, a Laser-Lok 3.0 implant design may be ideal for preserving a healthy status of bone, and *(3)* numerically demonstrated, placing the implant-abutment interface 0.5 mm below the bone crest might be an appropriate

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option to reduce the risk of overloading and subsequent bone loss. Additionally, placing the implant shoulder subcrestally has significant clinical advantages, eg, in esthetic sites to avoid the unesthetic appearance of the metal crown margin, especially for patients with a high smile line, and is favorable for an ideal emergence profile.

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