Impact Fracture Resistance of Two Titanium-Abutment Systems Versus a Single-Piece Ceramic Implant

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ABSTRACT

Background: The number of patients with oral implants has increased significantly. However, the literature addressing the effect of impact force on titanium and/or ceramic implants is inconclusive. This study sought to determine the fracture resistance to impact load of titanium and ceramic endosseous oral implants.

Materials and Methods: Endosseous oral implants were vertically positioned in two different mounting media: brass and a bone-simulation material. The implant configurations tested included an experimental one-piece Y-TZP implant and a commercially available titanium implant (external hex) with both titanium and zirconia abutments. The specimens were subjected to an impact load using a pendulum impact tester with tup weights varying from 0.9 to 4.5 kg delivered at a radius of 40.64 mm. Loads were delivered to the abutment at a point 4.27 mm above the implant fixture and block junction. Statistical differences ($p < .05$) were established using the F-test for variances and, when different, t-test assuming unequal variances.

Results: For implants clamped in brass, the titanium implant with titanium abutment required the greatest energy to fracture the implant-abutment system (only the abutment screw failed). The ceramic implant and ceramic abutment on titanium implant presented the lowest fracture energy ($p < .01$). No significant differences were observed when different systems were inserted into the foam blocks of the bone substitute ($p > .25$).

Conclusion: This investigation showed that the fracture energy of two titanium-abutment systems versus a single-piece Y-TZP implant in foam blocks simulating bone elastic modulus was not different, and that differences occurred when the embedding material elastic modulus was increased an order of magnitude.

KEY WORDS: ceramic, fracture resistance, impact load, oral implants, titanium

Implant-supported prostheses have been shown to be an effective oral rehabilitation treatment modality.[1,2] Because of its long-term outcome, pulpally involved teeth may be extracted and replaced with an implant rather than receiving conventional endodontic treatment. As a consequence, more and more adults are receiving implants to replace missing teeth.

Much has been documented about the risks of fracture to natural teeth, especially in children and adolescents. Also, considering the fact that today's active lifestyles of adults present similar risks, adults have become more prone to impact forces directed to their mouth. Between 1991 and 1998, of 784 patients aged 7 to 32 years with alpine ski-related injuries, 42% had

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tooth fractures.[3] Four percent of whitewater rafting injuries are to the teeth.[4] Injuries from accidents, altercations, and falls could cause tooth fracture. Of trauma-related dental injuries (e.g., automobile accidents, etc.) treated in after-hour emergency settings, nearly 30% were for fractured teeth.[5] As people age, there is an increase in the rate and absolute number of injuries, and the contribution of falls to trauma increases.[6]

The fracture resistance of natural teeth to impact load has been extensively documented, and the values for anterior teeth are presented in Table 1. However, the literature concerning the impact fracture resistance of endosseous oral implants is sparse and inconclusive. To address this question, the objective of this study was to determine the fracture resistance to impact load of a titanium implant with titanium and Y-TZP abutments versus a one-piece zirconia ceramic endosseous oral implant.

MATERIALS AND METHODS

The implant configurations tested included a commercially available external hex titanium implant (Brånemark System® Mk III Groovy RP 3.75 mm diameter by 13 mm length, Nobel Biocare, Göteborg, Sweden) with a titanium abutment (Brånemark System RP, 3 mm collar, Nobel Biocare) and with an esthetic zirconia abutment (Brånemark System, 1 mm collar, Nobel Biocare) and an experimental one-piece zirconia implant with a 4.3 mm diameter and 13 mm length.

The endosseous oral implants were positioned and tested in two different mounting media: in a brass sleeve (brass) and in a block of bone-simulating material

TABLE 1 Energy to Fracture Natural Anterior Teeth

Tooth/Condition	Energy to Fracture (J)	Reference
Maxillary central incisor – intact		
• Pseudo pdl*	0.355 ± 0.005	7
• Formalin + pseudo pdl	1.009 ± 0.282	8†
• Formalin, no pdl	0.346 ± 0.179	9
• No pdl	0.30 ± 0.08	10
Maxillary central incisor – restored		
• Vita Dur N ceramic crown	0.951 ± 0.432	8†
• Gold crown	0.901 ± 0.432	8†
• Vita Hi-Ceram crown	0.895 ± 0.353	8†
• Dicor glass-ceramic crown	0.561 ± 0.117	8†
• Vita Dur N veneer	0.957 ± 0.432	8†
• RCT‡ composite core + Pd-Ag crown with collar	0.295 ± 0.075	9
• RCT composite core + Pd-Ag crown without collar	0.159 ± 0.047	9
Mandibular incisor		
• Intact	0.29 ± 0.07	10
• RCT with gutta-percha fill	0.257 ± 0.174	11
• RCT with stainless steel post	0.351 ± 0.188	11
• RCT + carbon fiber post	0.247 ± 0.125	11
Maxillary lateral incisor		
• Intact	0.28 ± 0.11	10
Maxillary canine		
• Intact	1.26 ± 0.12	10

Values are given in joules for comparison across studies; 1 ft-lb = 1.356 J.

*pdl, periodontal ligament. Pseudo ligaments were created using a complaint layer such as silicone rubber, between the fixturing material and the tooth.

†See discussion above about why these results are substantially higher than those of other studies by the same[7] and by other groups.[9,10]

‡RCT, root canal treatment.

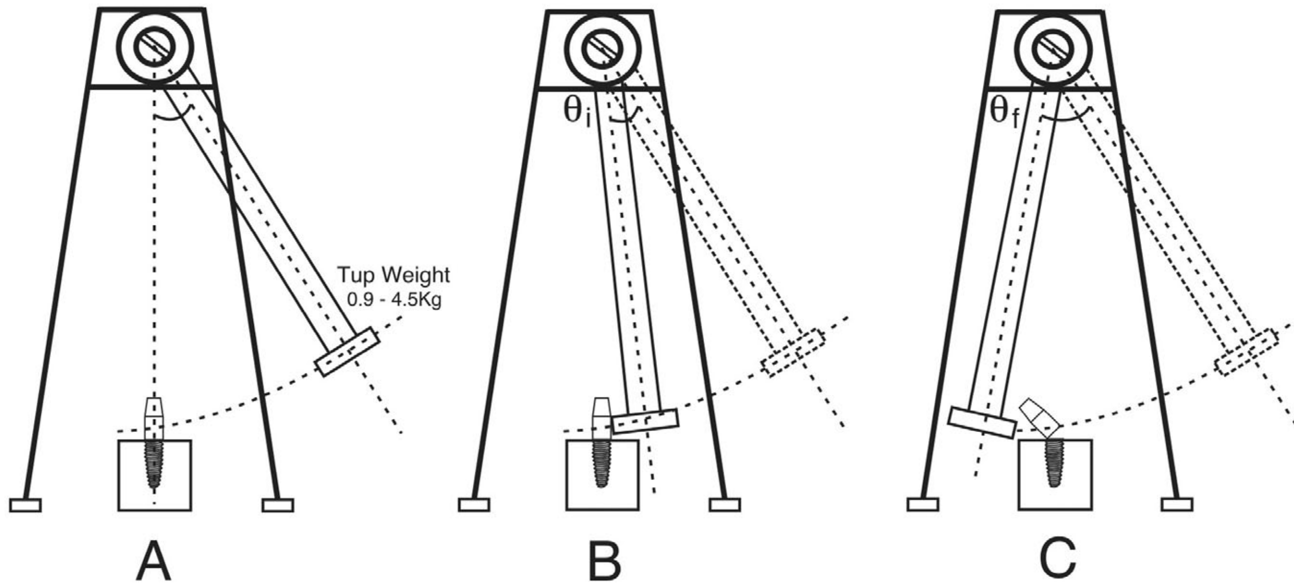

Figure 1 This figure represents a schematic of the impact test performed in this investigation. *A* shows the test setup. Note that the arm is released toward the implant fixed in a vise (B). The tup weight varied from 0.9 to 4.5 kg. *C* shows the implant fractured after impact. Note θ*i* and θ*f* showing the angle of release point and point of impact (B) and the angle between the release point and maximum follow-through (C), respectively.

(Sawbones, Pacific Research Laboratories, Vashon, WA, USA).[12] The brass sleeve was 4.34 mm inner diameter and 6.35 outer diameter and 25.4 mm in length. The implants of all groups were held in place with epoxy adhesive (Loctite® E-30CL, P/N 29329, Henkel Loctite, Rocky Hill, CT, USA). The bone simulation material was polyurethane foam (foam blocks) with density of 0.48 and 0.80 g/cm^3, and elastic modulus of 1.4 GPa (comparable to trabecular bone)[13] and 9.4 GPa (comparable to cortical bone),[13] respectively. The implants were vertically inserted into foam blocks with dimensions of $10 \times 15 \times 76$ mm. The implants were located in the block or brass with one thread exposed above the gripping surface.

After mounting, the specimens were subjected to an impact load (Figure 1) using a pendulum impact tester (TMI, Horsham, PA, USA) with tup weights varying from 0.9 to 4.5 kg delivered at a radius of 40.64 cm. Loads were delivered at a point 4.27 mm above the implants to foam block or brass clamp junction. The height of the pendulum before and after impact/fracture was digitally recorded as an angle. The reduction in angle following the implant impact allowed calculation of the energy absorbed during fracture. The starting and ending angle, and the constants comprising the impactor mass (m) and pendulum length (r) were used to calculate the fracture energy (Ea) according to the following equation:

$$\mathrm{Ea}\ (\mathrm{J}) = m \cdot r \cdot (\cos\theta f - \cos\theta i)$$

where θf is the angle between the release point and maximum follow-through of the pendulum, and θi is the angle of release point and point of impact.

Six groups were tested: titanium implant with a titanium abutment clamped in brass (TEB) ($n = 9$) and mounted in foam (TEF) ($n = 6$), titanium implant with zirconia abutment in brass (TZB) ($n = 10$) and in foam (TZF) ($n = 6$), and single-piece Y-TZP ceramic implant in brass (CB) ($n = 10$) and in foam (CF) ($n = 6$). Statistical differences ($p < .05$) were established using paired F-test for equal variances and paired t-tests assuming unequal variances.

RESULTS

Results are summarized in Table 2. For implants clamped in brass, the titanium implant with titanium abutment (group TEB) required the greatest load to fracture the implant-abutment system ($p < .01$), and then only the abutment screw failed (Figure 2A). The CB and TZB groups presented the lowest resistance to fracture when compared to all other groups ($p < .01$). In all ceramic implant specimens clamped in brass (group CS), the fracture occurred at the first thread of the implant (see Figure 2C). Catastrophic fracture of the abutment region was observed in all specimens of

TABLE 2 Fracture Resistance of Endosseous Implants and Fracture Location

Group	Number of Specimens	Load (kg)	Fracture Energy (J)	Fracture Location
Steel				
• TEB	9	4.5	21.18 + 3.05[a]	Abutment screw
• TZB	10	0.9	1.71 + 0.47[b]	Abutment fractured
• CB	10	0.9	0.75 + 0.09[b]	Stress riser at fixture point, failed at first thread (within fixture)
Foam block				
• TEF	6	4.5	4.89 + 1.00[c]	Foam fractured
• TZF	6	4.5	5.00 + 2.70[c]	Foam fractured (4), abutment fractured (2)
• CF	6	4.5	3.21 + 0.73[c]	Foam fractured

Values are given in joules for comparison across groups and studies; superscript letters in fracture energy indicate statistical differences ($p < .05$). Note that the different weights utilized for testing were accounted for during energy calculations.

the TZB group (see Figure 2E), and no damage on the abutment screw was observed after the impact test.

In the tests with the implants inserted into foam blocks, the fracture energy for all three test groups (TEF, TZF, and CF) were not statistically significant ($p > .25$), ranging from 3.2 to 5 J (see Table 2). In all but 2 of the 18 specimens tested, the implant remained intact and was displaced from the embedding material with failure of the embedding material (see Figure 2, B, D, and F). In two cases with zirconia abutments in foam blocks, the abutments fractured.

DISCUSSION

The present study sought to determine the fracture resistance to impact load of titanium implants with titanium and Y-TZP abutments versus one-piece ceramic endosseous oral implants. Our findings indicate that the mounting method employed has a dramatic effect on the nature of the impact results. Epoxy luting into a brass fixture leads to failures of the abutment screw for the titanium abutment, fracture of the zirconia esthetic abutment, and fracture of the one-piece ceramic implant. Mounting in bone simulating foam blocks resulted predominately in dislodgement of the implant from the block regardless of the bone density employed. The question remains as to whether the foam block mounting is representative of alveolar structure. A comparison to impact studies on natural teeth is warranted.

In the studies referenced in Table 1, the teeth are embedded in resin blocks sometimes with a periodontal ligament simulator created by adding a thin

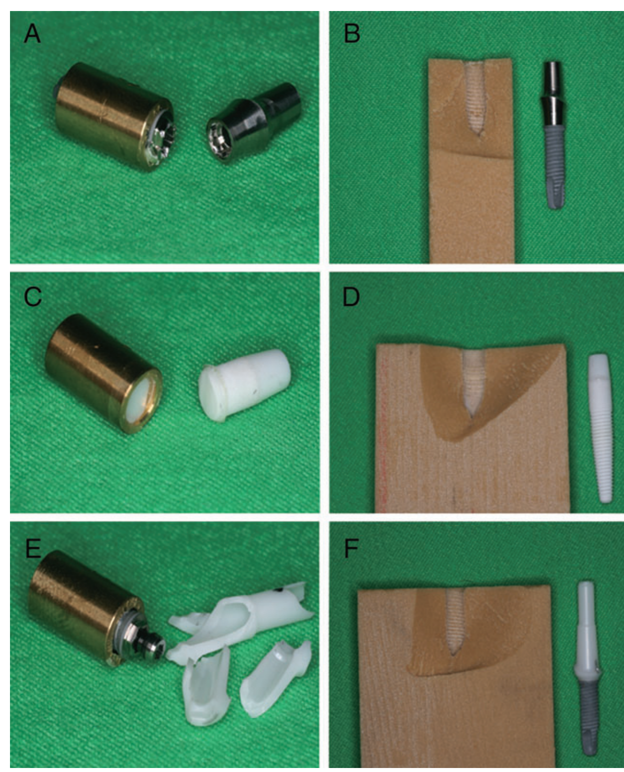

Figure 2 This picture shows representative images of fracture patterns of a group of specimens fixed in brass sleeves (images A, C, and E) and in foam blocks (images B, D, and F) after an impact test was performed. In *A*, the samples fractured near the implant hex. In *B*, the titanium implant was not damaged upon impact, but the foam blocks broke away with a fracture that split the foam across its width. Image *C* shows the ceramic implant bonded in the brass sleeve past the first thread. Fracture occurred at the first thread. In *D*, the ceramic implant was not damaged upon impact and foam blocks sheared at the edge that was held in the vise, similarly to *B*. *E*, The zirconia abutment fractured upon impact. The majority of failure condition for this group is shown in *F*, where a wide foam block piece broke after test.

compressible layer (e.g., silicone) between the tooth and the resin. The elastic modulus of the embedding resins utilized in testing the fracture energy of natural teeth is in the range of 1.6–3.0 Gpa, and is considered representative of the elastic modulus of the maxillary or mandibular alveolar bone complex.[14] The foam material utilized in the present study had an elastic modulus comparable to the resins and bone.[13] As documented by Andreasen and Andreasen[10] (see Table 1), adding the simulated periodontal ligament does not substantially affect the impact energy to fracture. However, the setup utilizing the brass sleeve represents a somewhat different scenario compared to the resin blocks utilized in the present and previous studies.

In all studies evaluated, the natural teeth fractured when impact loaded[8] with values ranging from 1.26 + 0.12 J for intact maxillary canines, to a low of 0.28 + 0.11 J for intact maxillary lateral incisors. However, the maxillary central incisors have been the focus of much attention and with one exception fractured when subjected to approximately 0.30–0.35 J of energy.[7–9,15] In contrast, the natural teeth values reported by Stokes and colleagues[15] were consistently higher compared to other studies.[9] A possible explanation for the differences may be that in one of the studies, the teeth were stored in formalin, possibly affecting the tooth micro- and macrostructure mechanical properties.[7]

Based on the studies reported in the literature, natural teeth are highly vulnerable to fracture when subjected to impact energies of approximately 1.26 J or less. Our results showed that implants mounted in bone substitute with similar properties to those materials used to test natural teeth (1.6–2.9 GPa for resins) can withstand substantially greater impact energies than natural teeth. In all groups, the impact fracture energy was nearly three times higher than the energy needed to fracture the most resistant natural teeth (maxillary cusps), and more than 10 times greater than the values needed to fracture intact maxillary central incisors (with one exception).[15]

An important difference between the fracture resistances measured is the fracture mechanism. In the studies of natural teeth, the tooth itself fractured. Continuing studies of tooth trauma by the American Academy of Pediatric Dentisty[11] suggests that teeth partially fracture (chip), totally fracture at various levels relative to the alveolar bone crest, or are avulsed. This is a fail safe system to maintain the alveolar structure. In the implant study, no implant, either titanium or ceramic, failed. In contrast, the implants were "avulsed" from the bone substitute (the foam fractured) or an abutment fractured (however, at energies more than 10 times greater than those required to fracture natural intact maxillary incisors). This suggests that when titanium of ceramic fixtures is osseointegrated, facial trauma is likely to lead to alveolar fracture leaving the implant intact. Also, the differences in fracture mechanism suggest that should a patient receive a blow that would fracture a natural tooth, it will not likely fracture an osseointegrated implant.

While no significant differences were observed regarding the fracture energy of the different implant setups in the foam blocks, changing the mechanical properties of the embedding material an order of magnitude (~100 GPa for brass vs 1.4 GPa for foam blocks)[16] resulted in significant differences between implant systems. Because bone mechanical properties in the jaws range from less than 1 GPa to approximately 30 Gpa,[17] further studies including materials simulating different bone mechanical properties should be encouraged.

CONCLUSION

This investigation showed that the fracture energy of zirconia- and titanium-abutment systems versus a single-piece Y-TZP implant in bone simulating foam blocks was not different, and that differences occurred when the embedding material elastic modulus increased an order of magnitude.

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