Resistance of Internal-Connection Implant Connectors Under Rotational Fatigue Loading

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Purpose: *To aid in developing mechanically optimized implant-abutment connectors, the fatigue resistance of 5 connector configurations of the Replace Select system (Easy abutment, Easy abutment without antirotational mechanism, Multi-unit abutment, Esthetic Alumina abutment, Esthetic Zirconia abutment) was investigated. Other purposes of the study were to determine whether the connector's antirotational mechanism participates in fatigue resistance and to compare the results with previous data on Straumann connectors.* ***Materials and Methods:*** *The repetitive, alternating, and multivectorial intraoral force pattern was reproduced by subjecting the test specimens to the rotating cantilever beam test. To this end, the samples were spun around their long axis while clamped into a revolving collet on one end and loaded normal to their long axis on the other end. The aim was to determine the load level at which 50% of the specimens survived and 50% fractured before 10^6 cycles. Means were determined using the staircase procedure. They were fitted with 95% confidence intervals for intergroup comparisons.* ***Results:*** *In the chosen testing configuration, 2 statistical groups emerged. The Easy abutments with and without antirotational mechanism were statistically similar, with mean failure loads in the 70 to 72 N range. Both ceramic and the Multi-unit abutments belonged to the second group, with mean failure loads in the 53 to 58 N range.* ***Conclusions:*** (1) *The fatigue resistance of ceramic and the Multi-unit abutments was approximately 20% less than that of the Easy Abutments.* (2) *The antirotational mechanism did not participate in mechanical resistance.* (3) *The fatigue strength of the Easy abutment connectors was approximately 20% greater than the equivalent abutments in the Straumann system.* INT J ORAL MAXILLOFAC IMPLANTS 2007;22:249–257

Key words: *connectors, dental implants, failure, fatigue*

Intraoral fracture of industrially machined implant parts is an infrequent but traumatic event for the patient and the clinician. All components that is, the implant, the connecting screws, and the abutment (if present), are susceptible to fracture.[1,2] Reported implant failure rates have been as low as 0.1%[3] or 0.6%[4] and as high as 2.7%[5] or 3.5%[6] (the latter was reported in an early study by Adell et al). A systematic review demonstrated an implant fracture rate of 0.4% at 5 years and 1.8% at 10 years.[7] With respect to screws, the heterogeneity in clinical applications (single-tooth restorations, multiunit restorations, fixed versus removable anchorage), test duration, and systems precludes the merging of reported data into overall breakage estimates. Nonetheless an extensive survey indicated a "mean incidence" of prosthesis screw fracture of 4% for prosthesis screws and 2% for abutment screws.[8] No screw fracture was reported for Straumann components in a recent clinical report spanning an 8- to 12-year period.[9]

Intraoral prosthetic components seldom fail after a single intense course of load application. Typically fracture occurs after a large number of chewing cycles, suggesting that as a small defect at the surface of a component gradually transforms into a crack, many thousands of load applications are required before the crack reaches its critical size and breakage ensues.[10] In addition, chewing imposes forces on teeth (or implant-supported prostheses)

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that have both a vertical and a horizontal component.[11,12] While the restoration's occlusoapical height has no bearing on the vertical component, it may significantly influence the horizontally directed forces that act on the connector. In effect, transverse forces are considered the most detrimental because of the inferior resistance of the components to tension or shear forces and the bending moment caused by crown height.[13,14]

Thus, a laboratory test should duplicate both the repetitive and the multivectorial nature of functional force application. By securing the sample at one end and applying a bending force to the other while the sample is rotated around its long axis, the test proposed herein subjects the samples to a 360-degree field of transverse tensile and compressive force vectors. The test was originally introduced in the mid-1800s by Wöhler for the development of new alloys for railroad axles.[15] It has been applied to analysis of the fatigue resistance of prosthodontic structures,[16,17] resinous materials,[18] and adhesive interfaces.[19]

Fatigue resistance of a component is commonly analyzed by developing an *S-N* diagram in which the ordinate *S* is the stress and the abscissa *N* the number of cycles sustained before breakage.[20] Typically such diagrams may be drawn for up to 10^7 to 10^8 cycles. For applications in which a predetermined cycle range is expected (as in prosthodontics), drawing full *S-N* diagrams may not be the most efficient approach. Instead one may arbitrarily set a number of cycles for which the component's fatigue strength will be determined.[21] Several techniques have been recommended for this type of analysis, which is based on quantal (fail or nonfail) data.[22] For a reliable estimate of fatigue strength, larger numbers of specimens are required. Probit (probability unit) analysis[23] is usually recognized as the procedure yielding the greatest accuracy. However, the procedure requires in excess of 50 specimens and substantial numerical treatment. Alternatively, the staircase technique[24] is a straightforward procedure in which a series of samples is tested consecutively. This technique has been used in the present study.

The goals of this study were *(1)* to compare 4 types of abutments (the Easy abutment, the Multi-unit abutment, the Esthetic Alumina abutment, and the Esthetic Zirconia abutment) for the Replace Select implant system (Nobel Biocare, Göteborg, Sweden), *(2)* to determine whether the connector's antirotational mechanism participates in fatigue resistance, and *(3)* to compare the present results with data gathered in a previous study on Straumann connectors.

MATERIALS AND METHODS

Overview

To duplicate the multivectorial force pattern of the mouth, the specimens (ie, the implant-abutment-restoration combinations) were configured as rotating cantilever beams. In this arrangement the inner end (ie, the implant) was clamped into a rotating collet, while the protruding end (ie, the restoration) was loaded perpendicular to the specimen's long axis. When the specimens were rotated, the connectors were subjected to sinusoidal tension-compression cycles which, depending on the magnitude of the load applied, caused breakage of the components. The goal of the experiment was to determine the force level at which 50% of the specimens would survive 10^6 cycles and 50% would fail.

The principle of the test required that the implants, abutments, and restoration analogs be collinear. Furthermore, the length of the lever acting between the collet and the point of force application had to be kept constant (Fig 1).

Technical aspects of the machinery and ancillary controls were described in previous reports.[25]

Implants

The objective of this study was to evaluate selected connectors for the Replace Select implant system (Nobel Biocare). To this effect, straight, regular-platform implants (part no. 28954) were chosen. The implants were coated with the company's proprietary electrodeposited oxide layer (TiUnite). They were 15 mm long, with a diameter of 4.3 mm. The connector design was characterized by an external flat surface normal to the implant's long axis. In the central portion of the connector, rotational stability was provided by 3 semicircular notches (ie, cams). More apically, the connector included a 4-mm-deep cylindric recess and an M2 screw tread. Figure 2 details the specifications of the implant and the female portion of the connector.

In rotational fatigue tests, the specimens are typically clamped into steel collets. However, this generates stress concentrations in the specimens at their interface with the grips and may cause undue (ie, not clinically relevant) breakage of the implant body. Therefore, before being clamped into the collet, the implants were encased in specially prepared aluminum tubes. A 0.5-mm-wide longitudinal slit was cut into the tubes to allow for some springiness during tightening. The load was applied to restoration analogs. These rotation-symmetric pieces were made of machined stainless steel and configured to allow accurate positioning to establish a 11.3-mm distance between the middle of the ball bearing and the

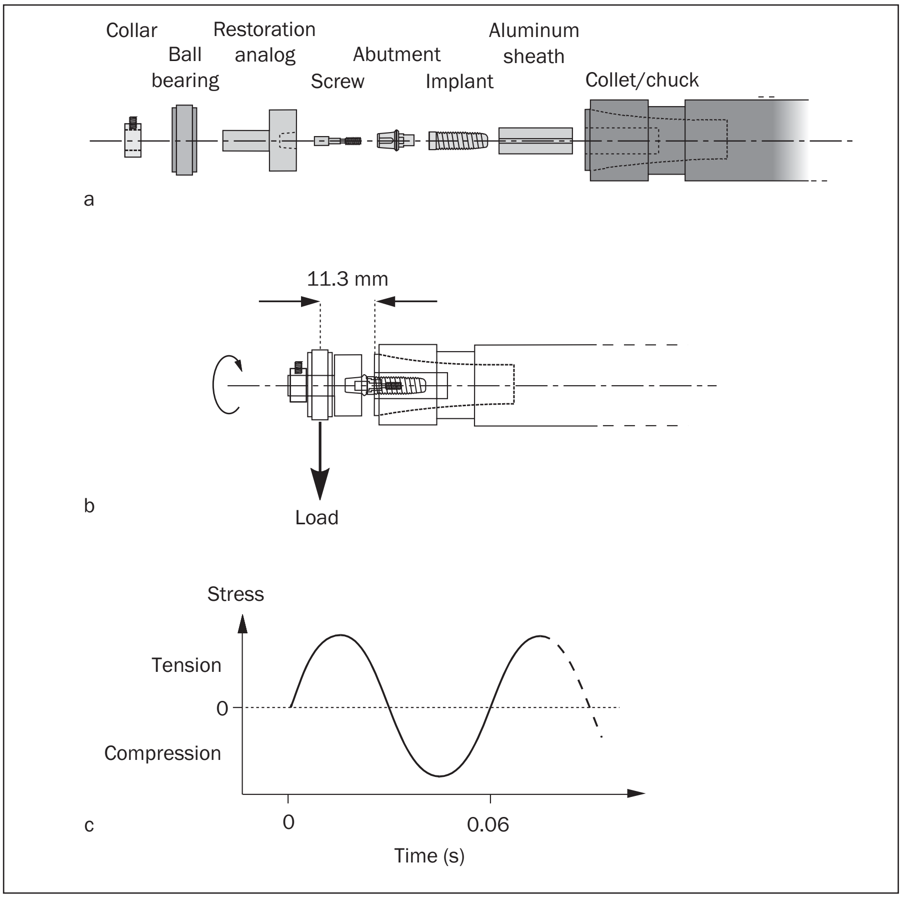

Fig 1 Testing setup for cantilever rotating beam fatigue test. *(a)* All parts were collinear and spun around their long axis. *(b)* The load was applied normal to the rotating specimen. The lever length was maintained constant at 11.3 mm. *(c)* Upon rotation, alternating sinusoidal tensile and compressive stresses developed inside the sample.

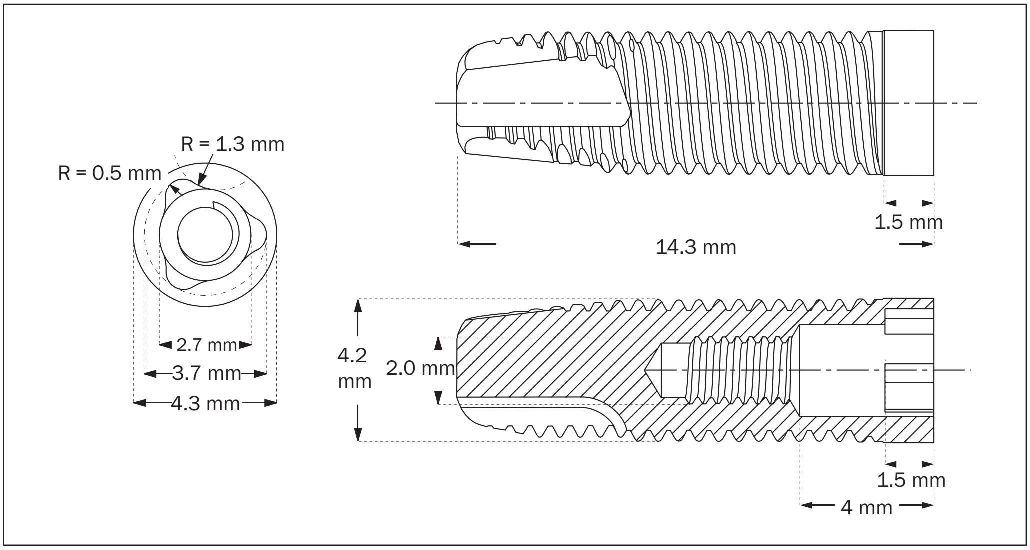

Fig 2 Dimensional specifications of the implant and the female part of the connector.

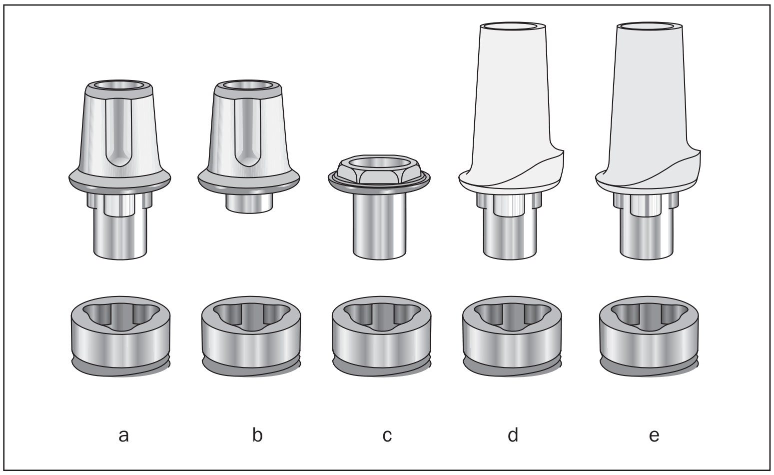

Fig 3 Tested Replace-Select abutment-connector configurations: the *(a)* Easy abutment, *(b)* Easy abutment without connector, *(c)* Multi-unit abutment, *(d)* Esthetic Alumina abutment, and *(e)* Esthetic Zirconia abutment.

collet (Fig 1). Typically the restoration analog was cemented to the abutment using composite cement (Variolink II; Ivoclar Vivadent, Schaan, Liechtenstein). For the Multi-unit abutment, it was machined so as to duplicate the system's gold cap geometry. It was then screw-fastened to the abutment using the M1.5 screw torqued to 15 Ncm.

Abutments

Five abutment configurations were evaluated. They are diagrammatically shown in Fig 3.

Easy Abutment. The Easy abutment (Fig 3a) is a conical titanium abutment with 3 longitudinal grooves. For the present tests, part no. 29471, a component with a height of 6.5 mm and a 1.5-mm collar, was used. The abutments were fastened to the implant with a surface-enhanced commercially pure titanium screw (TorqTite) tightened to 35 Ncm. In its apical portion, the abutment mirrored the female configuration of the connector, with 3 cams and a central bore for the screw. This abutment is typically used for cemented metal-ceramic fixed partial dentures (FPDs).

Easy Abutment with Antirotational Mechanism Removed. To test the hypothesis that the antirotational mechanisms do not participate in the overall mechanical resistance of the abutments, those mechanisms were carefully removed under magnification using an engineering lathe (Schaublin 102, Bevillard, Switzerland; Fig 3b). After the cams had been removed, the remaining tube was trimmed to a length of 1 mm. Full elimination was not possible, as the central tube provided the screw head's seat. To prevent any contact with the implant, the diameter of the tube was trimmed to 2.4 mm. The implants and abutment thus mated solely on the external surfaces while being centrally clamped by the titanium screws.

Multi-unit Abutment. The Multi-unit abutments (Fig 3c) used were 1-mm-high collar platforms with a hexagonal shape (part no. 29199). The screw that fastened the abutment to the implant was itself equipped with a M1.5 thread in its screw head. The restoration analog was configured so that the M1.5 screw clamped the analog onto the implant. Multi-unit abutments are typically used for screw-fastened restorations.

Esthetic Alumina Abutment. The Esthetic Alumina abutments (Fig 3d) used were 9-mm-high conical abutments intended for FPD cementation (part no. 29256). These are fitted with a metal insert that carries the cams and the central tube and stabilizes the ceramic cone relative to the implant. Theses parts are characterized by an asymmetrical apical collar. Fabricating the restoration analogs required special measures, as the ceramic cone's section is slightly oval and the cone's top is off-axis.

Esthetic Zirconia Abutment. These zirconium oxide-based components (part no. 30918; Fig 3e) are similar in design to the Esthetic Alumina abutments. The ceramic abutments are typically used in combination with full-ceramic restorations.

Experimental Procedure and Data Analysis

For each connector, the implant base and the length of the lever arm were kept constant at 11.3 mm. The sole varying parameter was the design and the material specifications of the screw-fastened connector. The experimental procedure required that a number of specimens be tested in a row. To this effect the specimens were loaded normal to their long axis via the ball bearing and spun at 1,000 rpm (16.7 Hz) for a maximum of 10^6 load cycles. After 10^6 cycles the test was halted, and the specimen was examined to determine whether it was broken or intact. If it had survived 10^6 cycles, the specimen was said to have "run out," and the next specimen was loaded to the previous magnitude plus 5 N. The same force (5 N) subtracted from the former load magni-

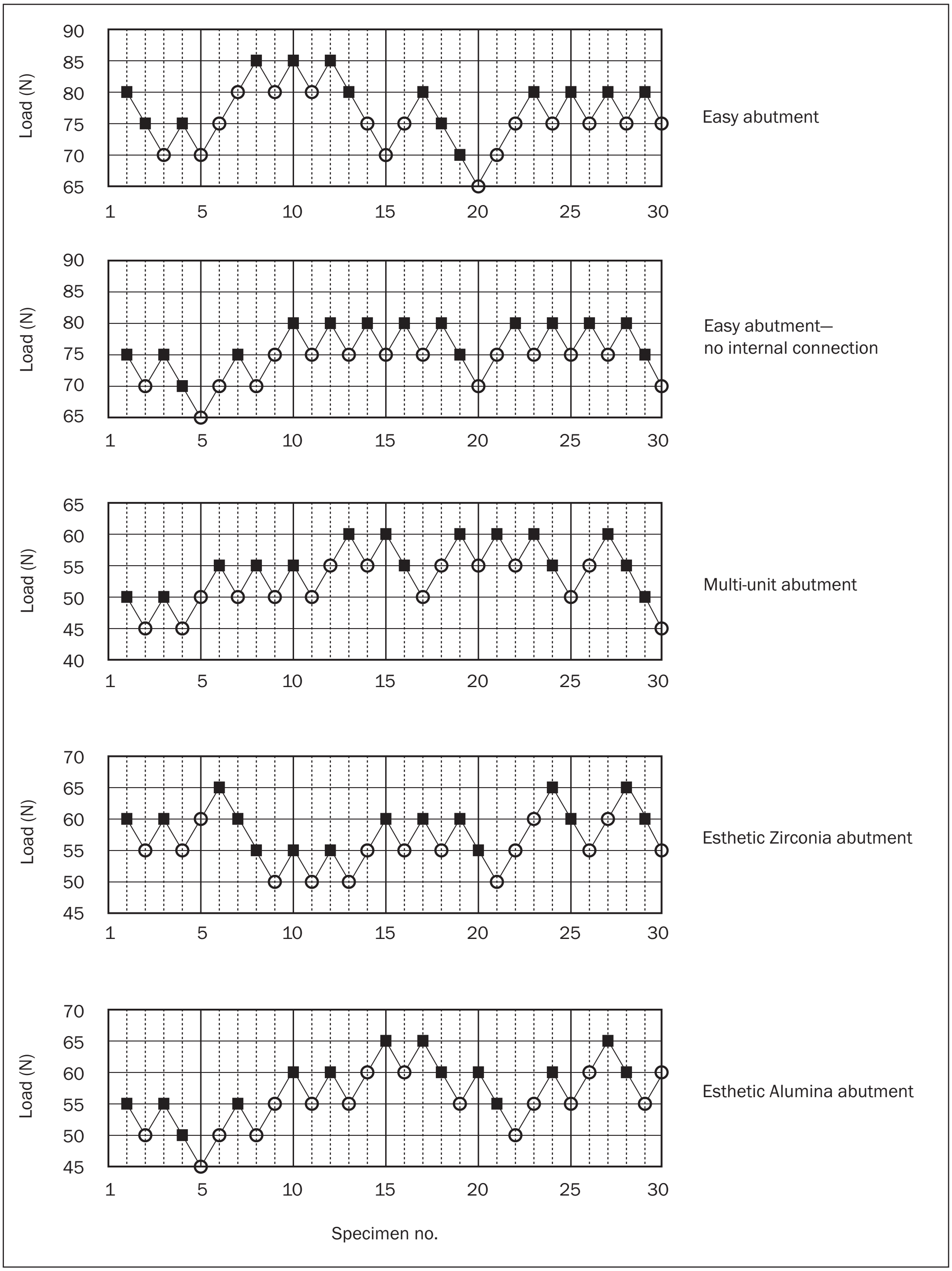

Fig 4 Staircase plots of the connectors tested.

tude if the previous specimen had failed. As the number of tested specimens increases, the characteristic "up-and-down" pattern for which the staircase procedure is named[26] develops. The goal of the procedure is to determine the load level at which 50% of the samples survive 10^6 cycles and 50% fail (F_{50}). In the present study, 30 samples were tested in sequence.

At the conclusion of the test, the results were first graphically arranged as in Fig 4. They were then tabulated as shown in Table 1, which then yielded the val-

Table 1 Example of Data Arrangement for Staircase Analysis (Easy Abutment)

Applied force in newtons	Force level (i)	No. of failures (n^i)	in_i	i^2n_i
80	3	3	9	27
75	2	8	16	32
70	1	3	3	3
65	0	1	0	0
		n = 15	A = 28	B = 62

n = Σn_i, A = Σin_i ; B = $\Sigma i^2 n_i$.

Table 2 Fatigue Resistance of the Connectors Subjected to the Rotating-Bending Test

			CI	
	Mean failure level	SD	Upper	Lower
Easy abutment	71.83	5.49	69.33	74.33
Easy abutment—no internal connection	70.17	3.33	68.31	72.02
Multi-unit abutment	53.50	4.77	51.21	55.79
Zirconia abutment	57.17	3.98	54.95	59.38
Alumina abutment	56.43	6.23	53.66	59.20

Mean failure level = force level at which 50% of the samples survive and 50% fail before 106 cycles. When connectors with overlapping CIs were combined, 2 groups were identified: *(1)* the Easy abutment with and without internal connection, and *(2)* the Multi-unit and ceramic abutments.

ues of A and B. If the number of runouts and failures differed, data analysis was based on the least frequent event. When F_0 (the lowest level at which failure occurred) was set to 65 N and F_{incr} to 5 N, F_{50} was calculated as follows:

$$F_{50} = F_0 + F_{incr}\left[\frac{A}{n} \pm \frac{1}{2}\right]$$

and the standard deviation was

$$SD = 1.62\ F_{incr}\left[\frac{nB - A^2}{n^2} + 0.029\right] \text{ if } \left[\frac{nB - A^2}{n^2}\right] \geq 0.3$$

where F_{50} was the mean force level at which 50% of specimens ran out and 50% failed, F_0 was the lowest load level at which failure occurred, n was Σn_i (n_i being the number of failures for each load level; see Table 1), A was Σin_i (i being load level), and B was $\Sigma i^2 n_i$.

For example, F_{50} was 71.8 N (SD 5.5) for Easy abutment (Table 1).

Statistical Analysis

To determine significant differences, the mean failure loads were fitted with 95% confidence intervals according to the technique described by Collins.[27] Means with overlapping intervals were considered equivalent.

More than 140 specimens were produced during the course of this experiment. Extra samples were required to adequately set the entry levels for the staircase procedure (ie, within the up-and-down boundaries) and to "calibrate" the specimens. Calibration was necessary so that the experimental model worked properly, that is, excluding specimen failure modes other than those that duplicated a clinically pertinent event. In 1 instance, the restoration analog broke loose from the abutment. This specimen was discarded, and a new specimen was used.

RESULTS

The mean stress levels at which 50% of the connectors survived 10^6 cycles and 50% failed, and their respective standard deviations and confidence intervals, are tabulated in Table 2. Means with overlapping confidence intervals belonged to the same population.

Statistically, 2 groups were identified: *(1)* the 2 Easy abutments and *(2)* the 2 ceramic abutments and the Multi-unit abutment. Abutments in group 1 had a mean failure level of about 70 to 72 N, while those in group 2 had a mean failure level of about 53 to 58 N.

None of the connectors loosened during the experiment. In group 1, the Easy abutment group, screw fracture caused the failure of the connector. The screw broke either at the boundary between the shank and the first thread or within the thread as it gripped the implant bore. In the Multi-unit specimens the M1.5 screws pulled loose from the M2 screw head. The ceramic abutment failed because of screw fracture, because the metal inset broke loose from the ceramic, or both. No ceramic abutments fractured.

DISCUSSION

Fatigue Resistance of the Components

When subjected to cyclic loadings below a threshold value, some materials (eg, steels) will not fail on any realistic timescale. When such a threshold exists, it is referred to as the material's "fatigue limit." Fatigue limits can be determined experimentally by running specimens through ever-increasing cycle numbers. Alternatively the experimenter may choose to determine the material's fatigue strength at a predetermined number of cycles. This value is referred to as the material's conventional "endurance limit." The procedure described herein was used to determine the connectors' endurance limit at 10^6 cycles. The complex geometry of the specimens, however, precludes the normalization of this data to stresses (as would be expected for a "true" endurance limit), and the force values gathered here are only valid within the present experimental configuration.

With respect to the fatigue resistance of the connectors, 2 groups emerged: *(1)* the Easy abutment group, which presented the highest fatigue resistance, and *(2)* a group comprising the Multi-unit abutment and the 2 ceramic abutments. The fatigue resistance of the latter group was inferior by approximately 20%. Hence the ceramic abutments were comparable to a fully metallic connector.

The present data demonstrated that the implant system's antirotational features have no bearing on the fatigue strength of the connector. This is understandable inasmuch as force transmission occurs via the surface normal to the clamping force resulting from the pretension of the screw, that is, the external "horizontal" surface of the connector.[28] The antirotational mechanism is oriented parallel to the clamping force; the inferior aspect of the male part never contacts the bottom of the implant's antirotational recess. As they are not preloaded, the "vertical" portions of the connector do not carry functional stresses. The Replace-Select coupling design therefore qualifies as a flat-to-flat connector (in contrast to the 16 degrees biconal geometry of the Straumann implant[29]). Thus, the antirotational feature should be considered an indexing mechanism whose purpose is to reproduce a chosen abutment position. Its participation in mechanical resistance is negligible.

In a previous study the fatigue strength of several ITI Straumann connectors was determined and compared.[30] Since the present tests were conducted using comparable implant diameters and identical levers, intergroup comparisons are possible. It appears that the ceramic abutments (torqued to 35 Ncm) of either system present comparable fatigue strengths (54.5 ± 2.3 N for Straumann, 57.2 ± 2.2 N and 56.4 ± 2.8 N for Replace-Select; means ± 95% CI). The Straumann Octa connector and the Replace Select Multi-unit abutment were also in the same range (58.8 ± 2.1 N for Straumann, 53.5 ± 2.3 N for Replace-Select). A large difference, however, was evidenced when plain conical abutments were compared. The fatigue strength of the Straumann standard abutment torqued to 35 Ncm was 55 ± 2.7 N, while that of the Replace-Select Easy abutment was 71.8 ± 2.5 N. This occurred in spite of the biconal coupling that typifies the Straumann connector; while the Replace-Select features a flat-to-flat external surface. One plausible explanation may be derived from the law of beams, which relates the stress inside the beam (*S*) to the magnitude of the applied force (*F*), the lever length (*l*), and the beam diameter (*d*) via the following equation[31]:

$$S = \frac{32\,Fl}{\pi d^3}$$

Hence a seemingly insignificant increase in diameter would lead to a considerably decreased stress magnitude inside the component. In the present situation, the maximum diameter of the Straumann cone is 3.5 mm, while the external diameter of the Replace-Select implant is 4.3 mm. If both components were treated as beams, this difference in diameter would yield a difference in internal stresses of 50%. Alternate explanations may be sought in the relationship between screw pretension, mating surface preload, and friction upon screw tightening as established for both types of connectors.

Testing Setup

The objective of the setup was to test the system's proprietary connector—not the intrabony portion of the implant or the restoration. In this respect, a fatigue test such as the one applied in the present study requires preliminary work to eliminate fracture sites that would be considered illegitimate (ie, not pertinent clinically). For instance, during initial tests, the implants systematically fractured at the tip of the collet grips, thus duplicating a situation in which all clinical implants would fracture at the bony emergence. After the latter design flaw had been eliminated by adding an aluminum sheath, the cement bond between the conical abutments and the restoration analog failed intermittently. Both failure sites were considered illegitimate, and the specimens' configuration was corrected accordingly.

In the present study, the samples were subjected to forces normal to their long axis (Fig 1): The testing setup did not subject the connectors to compressive or oblique forces. This approach is justified inasmuch as a material's resistance to shear is half to 1 order of magnitude less than its resistance under compressive loading.[32] Furthermore, the component's occlusogingival height has little if no bearing under compression, but the moment thus created is decisive in the survival of the part. Hence it is suggested that implant pillars are more likely to fail under transverse stress than under compression. In its present configuration, the testing setup does not detect screw loosening, but efforts are under way to duplicate this phenomenon using the rotating-beam test.

The present data did not demonstrate any difference in fatigue strength between connectors with or without the internal antirotational features. This finding contrasts sharply with a 1996 report by Binon in which the size of the internal abutment recess that matched the external hexagon on the test implants was gradually increased, thereby decreasing the fit between both mating parts.[33] Binon constructed a sophisticated machine that was capable of subjecting connectors to bending forces whose orientation was random but included the 360 degrees around the implant's long axis. Although the analysis of the fatigue data was somewhat rudimentary, the test showed a definite negative relationship between the size of the hex-recess gap and the resistance to fatigue loading. Binon's machine obeyed both requirements for fatigue testing of prosthodontic components in that loading was intermittent and the forces were multivectorial. In addition, unlike the present setup, this device also generated a compressive stress component. The origin of the difference between the 2 datasets is unclear. It might be hypothesized, however, that the magnitudes of internal strains during cyclic loading were contributory, as the author extensively discusses micromovements and abrasions occurring on the edges of the hex screw head. Other factors that could have affected the results are quality of fit of the mating interface, buildup and loss of screw preload, and loading amplitude.

Predictive Value of the Test

The predictive value of a laboratory test such as the one applied in the present study is of major importance if a correlation is to be established between clinical survival rates and laboratory data. To this effect, the approach used was to determine a prosthodontic configuration whose clinical survival rate is well established and may be considered optimal. This configuration was then used as a benchmark in the laboratory tests and in the evaluation of other configurations. This exercise may be attempted using plain conical abutments as the standard, which would translate into an endurance limit of approximately 60 N to 65 N in the present testing procedure. Applying this approach to ceramic abutments would entail that their life expectancy be comparable but marginally inferior to the benchmark. Indeed, recent clinical trials on such abutments indicate either no fracture[34,35] or a 2% to 8% fracture rate over the chosen observation period.[36,37] With due consideration of the location (anterior, posterior) and type of FPD (single, multiunit), a tentative correlation may be established between the laboratory tests and clinical survival rates.

ACKNOWLEDGMENTS

The authors' gratitude is expressed to Messrs P. Germann and C. Jaquet for their dedication during the machining of the ancillary parts of the specimens. They are also grateful to Mr W. Müller of Nobel Biocare (Switzerland) for providing the implant components.

REFERENCES

1. Luterbacher S, Fourmousis I, Lang NP, Bragger U. Fractured prosthetic abutments in osseointegrated implants: A technical complication to cope with. Clin Oral Implants Res 2000;11:163–170.
2. Tagger Green N, Machtei EE, Horwitz J, Peled M. Fracture of dental implants: Literature review and report of a case. Implant Dent 2002;11:137–143.
3. Buser D, Mericske-Stern R, Bernard JP, et al. Long-term evaluation of non-submerged ITI implants. Part 1: 8-year life table analysis of a prospective multi-center study with 2359 implants. Clin Oral Implants Res 1997;8:161–172.
4. Eckert SE, Meraw SJ, Cal E, Ow RK. Analysis of incidence and associated factors with fractured implants: A retrospective study. Int J Oral Maxillofac Implants 2000;15:662–667.
5. Naert I, Quirynen M, van Steenberghe D, Darius P. A study of 589 consecutive implants supporting complete fixed prostheses. Part II: Prosthetic aspects. J Prosthet Dent 1992;68:949–956.
6. Adell R, Lekholm U, Rockler B, Brånemark P-I. A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. Int J Oral Surg 1981;10:387–416.

7. Pjetursson BE, Tan K, Lang NP, Bragger U, Egger M, Zwahlen M. A systematic review of the survival and complication rates of fixed partial dentures (FPDs) after an observation period of at least 5 years. Clin Oral Implants Res 2004;15:625–642.
8. Goodacre CJ, Bernal G, Rungcharassaeng K, Kan JY. Clinical complications with implants and implant prostheses. J Prosthet Dent 2003;90:121–132.
9. Bragger U, Karoussis I, Persson R, Pjetursson B, Salvi G, Lang N. Technical and biological complications/failures with single crowns and fixed partial dentures on implants: A 10-year prospective cohort study. Clin Oral Implants Res 2005;16:326–334.
10. Wiskott HW, Nicholls JI, Belser UC. Stress fatigue: Basic principles and prosthodontic implications. Int J Prosthodont 1995;8: 105–116.
11. Graf H, Geering AH. Rationale for clinical application of different occlusal philosophies. Oral Sci Rev 1977;10:1–10.
12. van Eijden TM. Three-dimensional analyses of human bite-force magnitude and moment. Arch Oral Biol 1991;36:535–539.
13. Richter EJ. In vivo horizontal bending moments on implants. Int J Oral Maxillofac Implants 1998;13:232–244.
14. Genna F. On the effects of cyclic transversal forces on osseointegrated dental implants: Experimental and finite element shakedown analyses. Comput Methods Biomech Biomed Engin 2003;6:141–152.
15. Wöhler A. Versuche zur Ermittlung der auf die Eisenbahnwagen-Achsen einwirkenden Kräfte und die Wiederstandsfähigkeit der Wagen-Achsen. Zeitschrift für Bauwesen 1860; 10:583–616.
16. Libman WJ, Nicholls JI. Load fatigue of teeth restored with cast posts and cores and complete crowns. Int J Prosthodont 1995;8:155–161.
17. Basten CH, Nicholls JI, Daly CH, Taggart R. Load fatigue performance of two implant-abutment combinations. Int J Oral Maxillofac Implants 1996;11:522–528.
18. Scherrer SS, Wiskott AH, Coto-Hunziker V, Belser UC. Monotonic flexure and fatigue strength of composites for provisional and definitive restorations. J Prosthet Dent 2003;89:579–588.
19. De Munck J, Braem M, Wevers M, et al. Micro-rotary fatigue of tooth-biomaterial interfaces. Biomaterials 2005;26:1145–1153.
20. Klesnil M, Lukas P. Fatigue of Metallic Materials. Amsterdam: Elsevier Scientific, 1992.
21. Wiskott HW, Nicholls JI, Belser UC. Fatigue resistance of soldered joints: A methodological study. Dent Mater 1994;10:215–220.
22. Little RE, Jebe EH. Statistical Design of Fatigue Experiments. London: Applied Science Publishers, 1975.
23. Finney DJ. Probit Analysis: A Statistical Treatment of the Sigmoid Response Curve. Cambridge, UK: Cambridge University Press, 1952:236–245.
24. Dixon WJ, Mood AM. A method for obtaining and analyzing sensitivity data. J Amer Stat Assoc 1948;43:109–126.
25. Wiskott HW, Nicholls JI, Belser UC. The relationship between abutment taper and resistance of cemented crowns to dynamic loading. Int J Prosthodont 1996;9:117–139.
26. Maennig W. Statistical planning of fatigue tests. Int J Fracture 1975;11:123–129.
27. Collins JA. Failure of Materials in Mechanical Design: Analysis, Prediction, Prevention, ed 2. New York: Wiley, 1993:369–374.
28. Shigley JE, Mischke CR, Budynas R. Mechanical Engineering Design. Columbus, OH: McGraw-Hill, 2004.
29. Merz BR, Hunenbart S, Belser UC. Mechanics of the implant-abutment connection: An 8-degree taper compared to a butt joint connection. Int J Oral Maxillofac Implants 2000;15: 519–526.
30. Wiskott HW, Pavone AF, Scherrer SS, Renevey RR, Belser UC. Resistance of ITI implant connectors to multivectorial fatigue load application. Int J Prosthodont 2004;17:672–679.
31. Gere JM, Timoshenko SP. Mechanics of Materials. London: Chapman & Hall, 2000.
32. Craig RG, Powers JM. Restorative Dental Materials. St Louis, MO: Mosby, 2002.
33. Binon PP. The effect of implant/abutment hexagonal misfit on screw joint stability. Int J Prosthodont 1996;9:149–160.
34. Henriksson K, Jemt T. Evaluation of custom-made Procera ceramic abutments for single-implant tooth replacement: A prospective 1-year follow-up study. Int J Prosthodont 2003;16:626–630.
35. Glauser R, Sailer I, Wohlwend A, Studer S, Schibli M, Scharer P. Experimental zirconia abutments for implant-supported single-tooth restorations in esthetically demanding regions: 4-year results of a prospective clinical study. Int J Prosthodont 2004;17:285–290.
36. Andersson B, Taylor A, Lang BR, et al. Alumina ceramic implant abutments used for single-tooth replacement: A prospective 1- to 3-year multicenter study. Int J Prosthodont 2001;14:432–438.
37. Andersson B, Glauser R, Maglione M, Taylor A. Ceramic implant abutments for short-span FPDs: A prospective 5-year multicenter study. Int J Prosthodont 2003;16:640–646.