A Comparison of Three Mechanical Properties of Four Implant Designs

Jack I. Nicholls, PhD
Professor, Restorative Dentistry
University of Washington
Seattle, Washington

Christoph H.-J. Basten, DMD,
Dr med dent, MSD
Private Practice
Kirchberg
West Germany

Abstract

Four dental implant/abutment combinations were selected to determine: (1) the engagement length of the abutment connection; (2) the rotational tolerance of the abutment on the implant; and (3) the torque required to fail the attachment between the abutment and implant when the two were clamped together with the manufacturer's prescribed torque values. The engagement length was measured and recorded in microns. The rotational tolerance in degrees was calculated from coordinate measurements taken with the abutment rotated on the implant in the two extreme positions. The failure torque was recorded after the abutment had been fastened to the implant with the recommended torque value. A one-way analysis of variance (ANOVA) test disclosed the following. The Screw-Vent® combination demonstrated the longest abutment engagement length, followed in descending order by the Swede-Vent TL®, Nobelpharma, and Calcitek. Because of the interlocking connection between both the Swede-Vent TL® and Screw-Vent® components, these systems had zero rotational tolerance values. In contrast, the Nobelpharma and Calcitek systems had a mean rotational tolerance of 3.08 and 3.04 degrees respectively. There was no significant difference between the latter two systems. Swede-Vent TL® also showed the greatest torque required to fail the attachment, followed in decreasing order by the Screw-Vent®, Nobelpharma, and Calcitek. All torsional failures occurred as stripped corners of the male component, and in all cases, the abutment screw did not fail.

Learning Objectives

After reading this article the reader should be able to:
- distinguish between the various grades of titanium used for dental implants.
- discuss the advantages and disadvantages of various implant attachment connection features.
- understand the relationship between friction and clamping force in implant components.
- compare the manufacturer's recommended torque for tightening screws in fixtures.

Clinical and laboratory protocols relating to dental implants[1-5] and the need for a passive fit between the various components are well documented in the literature.[3,6-9] A passive fit minimizes stress on the prosthesis and its supporting components when the prosthesis is placed in function.

Prosthetic complications, however, have been related to component failure.[5,8-13] This type of clinical situation is demonstrated in Figures 1A through 1D. Figure 1A shows a four-unit mandibular bridge supported in vivo on two Grade 1 titanium implants, and Figure 1B shows radiographs depicting the level of osseointegration attained. After a period of 2 years in the mouth, the anterior implant failed, requiring removal of the bridge and the remaining segment of the fractured fixture. Figure 1C shows the bridge after it was removed from the mouth. The fractured segment of the anterior abutment is visible, and the fracture location in the implant was at the distal end of the abutment screw. Figure 1D shows a closer view of this implant fracture site. From the time that this bridge was in the mouth, it is safe to say that the implant failed as a result of load fatigue. The end

Figure 1A—Mandibular bridge supported on two Nobelpharma fixtures. The bridge spans from the first molar to the first premolar.

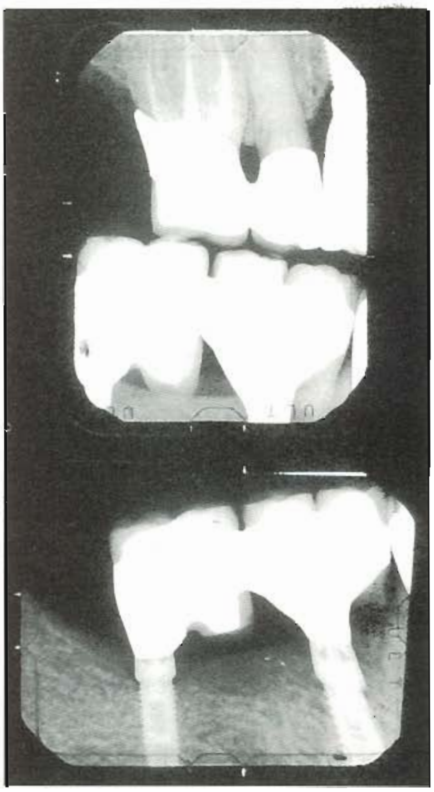

Figure 1B—X-ray films of fixtures in the mandible.

Figure 1C—Bridge removed from the mouth after fracture of the fixture below the premolars.

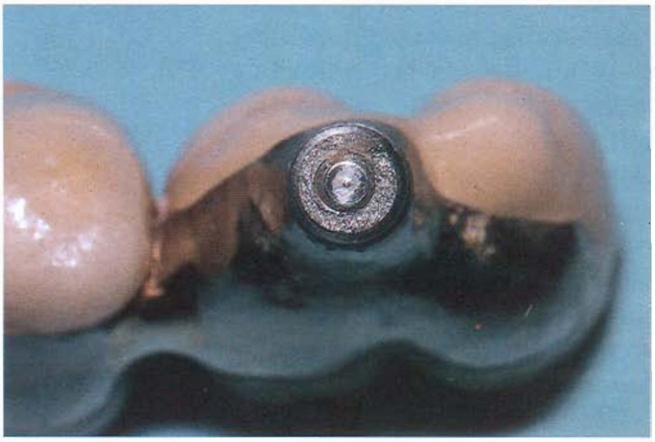

Figure 1D—Closer view of the fixture fracture surface. Fracture occurred at a level equivalent to the distal end of the abutment screw.

result is that the distal section of this anterior fixture must be removed surgically and, if possible, replaced. Binon[14] suggested that prosthetic screws can loosen as a result of: (1) inadequate tightening; (2) inadequate prosthesis fit; (3) poorly machined parts; (4) excessive loading; (5) screw design; and (6) elasticity of bone. Reasons three and five are related to engineering design, and thus can be assessed in a laboratory setting. Poorly machined parts, in particular, can be directly attributed to machining tolerances or a lack of machining accuracy. Balshi[15] also suggested that functional overload can lead to screw loosening and can also contribute to fixture, screw, and bone fractures, leading to a loss of osseointegration.

In an engineering sense, the machining tolerances of an implant's components are based on dimensional variation, surface roughness range of a component, and also the variation in mechanical properties that can result from heat treatment or other processing operations.[16] Both dimensional variation and surface roughness are types of machining tolerances.[16] The dimensional tolerance specifies how much a machined component can vary from an "exact dimension," before it fails or is defective. The surface roughness caused by machining also can affect an implant's tolerance because it can change the frictional force developed between the two surfaces in contact, ie, the friction between the abutment and the fixture.

The tolerances of implant components were discussed by Binon.[14] Implant components from several companies were measured, and their dimensional variations were recorded. In addition, the interface gap between components that were clamped together using standard screws torqued to the manufacturer's recommended values was also recorded.

Surprisingly, gaps existed between matched components from the same manufacturer, as well as between mixed components from different manufacturers. This could be significant because component settling or plastic flow of the contacting surfaces can occur in function with attendant loss of screw clamping force. Sorensen et al[17] also compared implant components from different manufacturers. The tolerance variations measured in their study were similar to those recorded by Binon.[14]

The purpose of the present investigation was to determine: (1) the rotational movement between the abutment/fixture attachment; (2) the length of engagement of this attachment; and (3) the torque required to fail this engagement length.

Figure 2—The four implant systems evaluated (shown from left to right): Nobelpharma, Dentsply Swede-Vent TL®, Calcitek, and Dentsply Screw-Vent®.

Table 1—Descriptions of Implant Systems Studied

Group 1	Nobelpharma: External hexagon
Fixture:	Commercially pure titanium—ASTM Grade 1
Abutment:	Commercially pure titanium—ASTM Grade 1
Group 2	Dentsply Swede-Vent TL®: External hexagon
Fixture:	Commercially pure titanium—ASTM Grade 4
Abutment:	Titanium alloy—ASTM Grade 23
Group 3	Dentsply Screw-Vent®: Internal hexagon
Fixture:	Titanium alloy—ASTM Grade 23
Abutment:	Titanium alloy—ASTM Grade 23
Group 4	Calcitek: Internal octagon
Fixture:	Titanium alloy—ASTM Grade 5
Abutment:	Titanium alloy—ASTM Grade 5

Methods and Materials

The four implant systems investigated were Nobelpharma[a], Dentsply Swede-Vent TL®[,b], Calcitek[c], and Dentsply Screw-Vent®[,b] (Figure 2, Table 1). Three separate test protocols were used to determine the three mechanical properties being evaluated in this study. Ten abutments in sealed vials were randomly paired with 10 fixtures from the same implant system. These 10 pairs were maintained during all measurements and testing.

Engagement Length Measurement

For this study, the engagement length was defined as the length of the male portion that would be in intimate contact when the components were assembled. In particular, this engagement length is that section of the attachment that would fail when

---

[a] Nobelpharma USA, Chicago, IL 60632
[b] Dentsply Implant, Encino, CA 91436
[c] Calcitek Inc, Carlsbad, CA 92008

Figure 3—Chatillon Torque Gage consisting of the calibrated wrench attached to the control and display unit.

Table 2—Torque Values Used to Tighten the Abutment Crown

Implant System	Tightening Torgue (N-cm)
Dentsply Swede-Vent TL®	30
Dentsply Screw-Vent®	30
Nobelpharma	20
Calcitek	22.6

the abutment was rotated with respect to the fixture.

Length measurements were made using a model 20 Nikon Measurescope[d] with an attached SC-102 digital counter. The accuracy of this device is 0.001 mm, and the reading error was found to be ±0.003 mm.

One edge of each fixture was aligned along one axis of the reticule in the eyepiece. This allowed direct measurements of the engagement length. Small angular misalignments ranged from 1 to 3 degrees. This would impose a maximum error of 0.14%.

Rotational Tolerance Between Abutment and Fixture

The objective here was to measure the amount of rotational movement between the abutment and the fixture, where this movement is directly related to the machining tolerances. These rotational movements were made using the model 20 Nikon Measurescope. First the abutment was attached to the fixture, and this combination was placed in a special measuring jig, which held the test components in a fixed vertical orientation. In this jig, the fixure was clamped against rotational movement using a set screw.

The steps required to calculate the rotational movement were:

1. First the abutment was rotated with respect to the fixture to the extreme clockwise position allowed by the machining tolerances.
2. The X and Y coordinates of a mark on top of the abutment were measured and recorded.
3. Next the abutment was rotated to the extreme counterclockwise position allowed by the machining tolerances.
4. The new X and Y coordinates of the mark on top of the abutment were measured and recorded.
5. Calculation of the rotational movement was accomplished using the two coordinate measurement sets.

The software used to make all of the mathematical calculations was Excel[e] version 5.0.

Torsional Failure of the Abutment/Fixture Combinations

To accomplish this, a Chatillon Torque Gage[f] was used (Figure 3). This precision instrument was calibrated to be accurate to within 5% of the scale value. It had the distinct advantage of being able to store and display the maximum torque value attained after a torque-loading test. The steps for determining the maximum torque to failure were:

1. The abutment was attached to the implant, and the abutment screw was torqued to the manufacturer's prescribed torque value (Table 2).
2. The implant was secured in the three-jaw chuck of the Chatillon Torque Gage.
3. The abutment was clamped to a small bench vice, allowing the torque gauge and the specimen to be held in a verti-

---

[d] Nikon, Tokyo, Japan

[e] Microsoft Corporation, Redmond, WA 98052

[f] Chatillon, Greensboro, NC 27409

cal orientation.

4. With the hand resting on top of the bench vice, the Chatillon Torque Gage was rotated clockwise until failure occurred. (By allowing the Torque Gage to be in a vertical orientation during this test, lateral movement of the gauge was not a problem because the hand rested on the vice top during torquing.)
5. The failure torque measurement displayed on the Chatillon Torque Gage was recorded.

Statistical Analysis

A one-way ANOVA test was used to determine differences between the four groups, and the Student-Newman-Keuls test identified statistically significant subsets at the 95% confidence level.

Results

The results of the three tests are given in Tables 3 through 5. In addition to the raw data, these tables give the mean and standard deviation for each implant system. Table 6 shows the results of the statistical analysis for each of the 3 variables measured.

Engagement Length of Implant Abutment Contacting Elements

Table 3 provides the engagement length measurements for each of the 10 attachments measured. The values in this table are in micrometers. The last 2 lines in this table give the means and standard deviations for the 10 samples in each of the 4 groups tested. The one-way ANOVA procedure determined 4 significant subsets (Table 6). Each of these subsets contained one implant system. The Screw-Vent® had the largest abutment engagement, followed by Swede-Vent TL®, Nobelpharma, and Calcitek, in decreasing order.

Table 3—Engagement Length for Each Implant System

Specimen Number	Engagement Length (μm) Nobelpharma	Dentsply Screw-Vent®	Dentsply Swede-Vent TL®	Calcitek
1	547	1456	618	540
2	531	1431	613	540
3	565	1487	635	529
4	567	1469	611	539
5	546	1455	615	516
6	560	1148	632	549
7	557	1454	628	532
8	542	1467	622	526
9	532	1469	607	545
10	562	1475	610	561
Mean	550.9	1461.1	619.1	537.7
Standard Deviation	13.2	15.7	9.8	12.7

Rotational Tolerance Between Abutment and Fixture

Table 4 gives the individual values of rotational tolerance for each of the samples tested in the four test groups. The means and standard deviations for these data are given in the last two lines of the table. The Swede-Vent TL® and Screw-Vent® had zero rotational tolerance values for all 10 samples However, the Calcitek and the Nobelpharma systems displayed mean rotational tolerance values of 3.08 and 3.04 degrees respectively. There was no significant difference between the rotational values for the latter 2 systems (Tables 4 and 6).

Torsion Failure of the Abutment/Fixture Combinations

The torsion failure values, plus the means and standard deviations, for the four systems tested are listed in Table 5. The statistical analysis disclosed 4 significant subsets for this variable, each subset containing 1 system (Table 6). According to this analysis, the Swede-Vent TL® system had the highest resistance to torsional force, followed by Screw-Vent®, Nobelpharma, and Calcitek, in decreasing order.

Discussion

Before discussing the results of this investigation, it may be helpful to explain information related

Table 4—Rotational Abutment Tolerance for Each Implant System

Specimen	Rotational Tolerance (Degrees)			
Number	Nobelpharma	Dentsply Screw-Vent®	Dentsply Swede-Vent TL®	Calcitek
1	3.4	0.0*	0.0*	3.3
2	2.5	0.0	0.0	3.5
3	2.2	0.0	0.0	1.9
4	1.1	0.0	0.0	1.4
5	3.8	0.0	0.0	3.6
6	2.4	0.0	0.0	3.1
7	3.8	0.0	0.0	3.3
8	3.8	0.0	0.0	3.9
9	2.3	0.0	0.0	3.6
10	5.5	0.0	0.0	2.8
Mean	3.08	**	**	3.04
Standard Deviation	0.08	**	**	0.08

* No discernable rotation under 25x
** Values not calculated

to ASTM specification B348-94 Standard Specification for Titanium and Titanium Alloy Bars and Billets. While this specification covers 23 grades of titanium, only 6 grades are used in dentistry. Grades 1 through 4 are for unalloyed, commercially pure titanium, whereas Grades 5 and 23 are for a titanium alloy containing approximately 6% aluminum and 4% vanadium. These grades are defined further in Table 7. The tensile strengths, yield strengths, and elongation of these 6 titanium grades are given in Table 8. The tensile and yield strengths also increase as the grade number increases. However, the elongation of these four grades decreases with increasing grade number. For the 2 titanium alloys (Grades 5 and 23), the tensile and yield strengths also increase with an increase in grade number. Grade 5 titanium alloy, therefore, has a higher contaminant level and higher tensile and yield values. The elongation, however, is the same for both Grades 5 and 23. It is clear then that higher grades of titanium are used for dental implants because the strength of the in vivo components is a major consideration.

The implant components tested in this protocol fit in the following categories:

Nobelpharma	Grade 1
Swede-Vent TL®	Grade 4
Screw-Vent®	Grade 23
Calcitek	Grade 5

Table 7 shows that while Grade 23 titanium alloy has the same basic components as Grade 5, its maximum allowable levels of carbon, iron, and oxygen are lower. Thus, Grade 23 is quoted as having "extra low interstitial elements" (ELI). This ELI designation indicates a lower contamination level and is purely a function of ASTM standards. This is not related to dental considerations.

On the basis of the titanium grade, it would be expected that Grades 4, 5, and 23 have a higher torsional strength than Grade 1. The tensile strength of Grades 5 and 23 is more than 3 times that of Grade 1, and the tensile strength in Grade 4 is twice that of Grade 1. Thus, the Screw-Vent®, Swede-Vent TL®, and Calcitek systems should have the highest torque values. However, the torsional failure values for the Nobelpharma components in this study were one to two times lower than the Swede-Vent TL®, Screw-Vent®, or Calcitek systems, rather than two to three times lower, as would be expected. This can only lead to the conclusion that the engagement length is not the only resistance invoked during torsional failure. Perhaps a second and significant resisting factor is the friction developed between the abutment and fixture. This factor alone has clinical significance when considered in the context of screw loosening.

There are two frictional forces to be considered here—static friction and dynamic friction. Static friction must be overcome to *induce* motion. Dynamic friction must be overcome to *maintain* motion. Static friction is always greater than dynamic friction. Thus, when the Nobelpharma and Calcitek systems are under the torsional load, static friction must first be overcome before motion occurs. This motion involves a slipping between the abutment and the fixture that is permitted as a result of the rotational tolerance (Table 4). As a result, the frictional force changes from static to dynamic at the point where the engagement length is activated. After the rotational tolerance has been eliminated, this engagement length is sheared. Thus, the final torsional resistance is a combination of dynamic friction and shearing of the engagement length. In contrast, because there was no rotational tolerance for the two Dentsply systems, torsional resistance would probably be caused by a combination of static friction plus shearing of the engagement length. The Screw-Vent® system has an engagement length at least twice that of the other three systems (Table 3). The relative failure torques (Table 5), however, do not reflect this same ratio. Therefore, it seems likely that a frictional component, whether static or dynamic, offers considerable resistance to torsional failure.

Table 5—Torsion Failure Values for Each Implant System

Specimen Number	Torsion Failure Values (N-cm) Nobelpharma	Dentsply Screw-Vent®	Dentsply Swede-Vent TL®	Calcitek
1	123.2	169.0	224.8	97.8
2	143.6	192.8	201.4	112.4
3	117.2	170.0	179.6	98.6
4	148.0	195.8	169.6	104.7
5	120.8	129.8	214.2	104.2
6	125.6	153.6	173.2	87.4
7	139.6	171.8	199.6	94.8
8	131.0	164.6	164.8	100.4
9	109.2	136.6	155.4	111.2
10	138.4	173.8	190.2	103.6
Mean	128.8	165.8	187.3	101.5
Standard Deviation	11.7	21.2	22.6	7.5

Table 6—Results of Statistical Analysis

	Statistical Subsets ($P \leq 0.05$) 1	2	3	4
Engagement Length	SC	SW	N	C
Rotational Tolerance Values**	N,C	-	-	-
Torsional Failure Values	SW	SC	N	C

* SW = Swede-Vent TL®, SC = Screw-Vent®, N = Nobelpharma, C = Calcitek.
** Only N and C were evaluated.

This is further exemplified by the fact that the Swede-Vent TL® system has an abutment engagement length and tensile strength comparable to those of the Calcitek system, yet the mean torsional failure value for the Swede-Vent TL® system is approximately 50% higher than that for the Calcitek system (Table 5). The frictional force developed between two clamped surfaces is a function of the clamping force times the coefficient of friction. It would appear reasonable to assume that the clamping force for the Swede-Vent TL® system was approximately 50% higher than that for the Calcitek system, and thus, the 50% difference between the failure torques becomes readily apparent.

This influence of friction must also be recognized in screw loosening. Screw loosening can be attributed to a settling of the components in some form of plastic flow or permanent surface deformation of the mating surfaces.[18] Jorneus et al measured the torque needed to overcome friction between the abutment and fixture for various tightening torques.[18] Their study showed that after settling, the torque decreased. Unquestionably, the torque delivery system can play a significant role here as well. Recent unpublished research by Tan and

Table 7—Grades of Titanium from ASTM B348-94

Grade Number	1	2	3	4	5	23
Nitrogen (max)	0.03	0.03	0.05	0.05	0.05	0.05
Carbon (max)	0.10	0.10	0.10	0.10	0.10	0.08
Hydrogen (max)	0.013	0.013	0.013	0.013	0.013	0.013
Iron (max)	0.2	0.3	0.3	0.3	0.4	0.25
Oxygen (max)	0.18	0.25	0.35	0.4	0.2	0.13
Residuals (max)	0.4	0.4	0.4	0.4	0.4	0.4
Aluminum	0.0	0.0	0.0	0.0	5.5-6.75	5.5-6.75
Vanadium	0.0	0.0	0.0	0.0	3.5-4.5	3.5-4.5
Titanium	*********** Remainder of Composition ***********					

Table 8—Tensile Strength, Yield Strength, and Elongation Values of Titanium Grades 1 Through 23

Grade	Tensile Strength (MPa)	Yield Strength (MPa)	Elongation (%)
1	240	170	24
2	345	275	20
3	450	380	18
4	550	483	15
5	895	828	10
23	828	759	10

Nicholls has shown that torque drivers may not be as accurate as proposed.

An additional factor related to this torque failure is the material from which the components are fabricated. In this respect, the coefficient of friction between the contacting metal surfaces is significant. For example, the engagement of the antirotational components, as well as the rotational tolerance, are very close in the Nobelpharma and Calcitek systems (Tables 3 and 4). However, this study revealed a significant difference between their torsional failure values (Table 5) with the failure in torque for the Nobelpharma system approximately 30% higher than that for the Calcitek system.

The torque values required to fail the components in torsion needs further clarification. One must ask whether these failure torque values would be attainable in vivo. The answer here can be derived from the works of Kydd,[19] Carr,[20] and Johansson.[21] These authors evaluated the torque required to rotate a fixture in bone. Kydd measured a maximum value of approximately 30 N-cm when a titanium fixture was implanted in a dog mandible. Carr found a torque value of 74 N-cm for commercially pure titanium, and 78.6 N-cm for a titanium alloy when the fixtures were implanted in a baboon jaw. Johansson recorded a value of 24.9 N-cm for commercially pure titanium in a rat tibia. These values are nowhere as large as the failure torques found in this study (Table 5). Thus, it is highly unlikely that torques of the magnitude given in Table 5 would be developed in vivo between the abutment and the fixture because the fixture would rotate before this torque magnitude was reached. It should also be mentioned that the rotational values would vary depending on the quality of bone and the size and design of the implants used.

From an engineering standpoint, the most important factor involved with joints held together with threaded members is the load induced in the shank of these threaded members by the tightening torque. The shank is the portion of the bolt or screw between the head and the threaded contact. Ideally, all of the torque applied to the abutment screw head should be resisted by the load in the shank of the bolt. Practically, however, this is not the case. Friction between the matching threads, and also friction between the head of the screw and the abutment, "reduce" the final shank load. Taken to an extreme, if there were no load in the shank of the bolt, no friction would develop at the abutment/fixture contact area. Thus, the load in the shank is highly important.

Conclusions

Four commercially available abutment/fixture systems were evaluated for length of engagement, rotational tolerance, and torsional failure. The results indicated the following:

1. The antirotational engagement lengths of each system were significantly different from one another. The Screw-Vent® system had the longest engagement (1.46 mm), followed by Swede-Vent TL® (0.62 mm), Nobelpharma (0.55 mm), and Calcitek (0.54 mm).
2. The rotational tolerance was zero for the Screw-Vent® and Swede-Vent TL® systems. The values for Nobelpharma and Calcitek systems were 3.08 and 3.04 degrees respectively. These tolerance values were not significantly different.
3. The torsional failure values for the four systems were all significantly different from each other. The highest resistance to torsional force was exhibited by the Swede-Vent TL® system (187.3 N-cm), followed by Screw-Vent® (165.8 N-cm), Nobelpharma (128.8 N-cm), and Calcitek (101.5 N-cm).

References

1. Brånemark P-I, Hansson BO, Adell R, et al: Osseointegrated implants in the treatment of the edentulous jaw: experience from a 10-year period. *Scand J Plast Reconstr Surg* 11(suppl 16), 1977.
2. Adell R, Andersson C, Brånemark P-I, et al: Manual of treatment with jawbone anchored bridges according to the osseointegration method. Gothenburg, Sweden, Faculty of Dentistry, Gothenburg, Sweden and the Institute for Applied Biotechnology.
3. Brånemark P-I, Zarb GA, Albrektsson T: *Tissue-integrated Prostheses: Osseointegration in Clinical Dentistry*. Chicago, Quintessence Publishing Co, 1985.
4. Lundqvist S, Carlsson GE: Maxillary fixed prostheses on osseointegrated dental implants. *J Prosthet Dent* 50:262-270, 1983.
5. Lindqvist LW, Carlsson GE, Glantz PO: Rehabilitation of the edentulous mandible with a tissue-integrated fixed prosthesis: a 6-year longitudinal study. *Quintessence Int* 18:89-96, 1987.
6. Zarb GA, Symington JM: Osseointegrated dental implants: preliminary report on a replication study. *J Prosthet Dent* 50:271-276, 1983.
7. Laney WR, Tolman DE, Keller EE, et al: *Dental Implants: Tissue-integrated Prosthesis Utilizing the Osseointegration Concept*. Mayo Clinic Proceedings 61:91-97, 1986.
8. Worthington P, Bolender CL, Taylor TD: The Swedish system of osseointegrated implants: problems and complications encountered during a 4-year trial period. *Int J Maxillofac Implants* 2:77-84, 1987.
9. Sones AD: Complications with osseointegrated implants. *J Prosthet Dent* 62:581-585, 1989.
10. Adell R, Lekholm U, Rockler B, et al: A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. *Int J Oral Surg* 10:387-416, 1981.
11. Cox JF, Zarb GA: The longitudinal clinical efficacy of osseointegrated dental implants: a 3-year report. *Int J Oral Maxillofac Implants* 2:91-100, 1987.
12. Adell R, Eriksson B, Lekholm U, et al: A long-term follow-up of osseointegrated implants in the treatment of totally edentulous jaws. *Int J Oral Maxillofac Implants* 5:347-359, 1990.
13. Zarb GA, Schmitt A: The longitudinal clinical effectiveness of osseointegrated implants: the Toronto study. Part III: problems and complications encountered. *J Prosthet Dent* 64:185-194, 1990.
14. Binon P, Weir D, Watanabe, et al: Implant component compatibility. In Laney WR, Tolman DE (eds): *Tissue Integration in Oral and Maxillofacial Reconstruction*. Chicago, Quintessence Publishing Co, 1990.
15. Balshi T: The role of screws in implant systems, Topic V in Special Supplement. *Int J Oral Maxillofac Implants* 9:48-63, 1994.
16. Shigley JE: *Mechanical Engineering Design*, ed 3. New York, McGraw-Hill, 1977.
17. Sorensen JA, Avera SP, Tomas C: Comparison of interface fidelity of implant systems. *J Dent Res* 70:540, 1991. Abstract.
18. Jorneus L, Jemt T, Carlsson L: Loads and designs of screw jonts for single crowns supported by osseointegrated implants. *J Oral Maxillofac Implants* 7:353-359, 1992.
19. Kydd WL, Daly CH: Bone titanium implant response to mechanical stress. *J Prosthet Dent* 35:567-571, 1976.
20. Carr AB, Larsen PA, Papazoglou E, et al: Reverse torque failure of various screw design implants in baboons. *J Dent Res* 73:1994. Abstract 804.
21. Johansson CB, Sennerby L, Albrektsson T: A removal torque and histomorphometric study of bone tissue reactions to commercially pure titanium and vitallium implants. *J Oral Maxillofac Implants* 6:437-441, 1991.

A958, rev. 10/07

This issue of ***Postgraduate Dentistry*** *has been made possible through an unrestricted educational grant from Dentsply Implant. The opinions expressed herein are those of the author(s) and do not necessarily reflect those of the editorial staff or the publisher.*