Theoretical cantilever lengths versus clinical variables in fifty-five clinical cases

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Statement of problem. Cantilever loading increases loads distributed to implants, potentially causing biomechanical complications. The implemented length is often less than what is considered to be optimal.
Purpose. This study investigated the effects of clinical variables on predicted cantilever lengths. Theoretically, calculated maximum cantilever was defined as the length that would not cause gold screw loosening or fatigue failure. The variables investigated included number and distribution of implants, arches placed, and the clinician's "optimal" cantilevers.
Material and methods. Implant and prosthesis location coordinates of 55 clinical cases were determined from casts. The distribution of an applied 143 N vertical load to implants was calculated through the Skalak model for more than 500 loading sites. Gold screw joint overload was assumed to occur at 200 and 250 N in compression and tension. Calculated lengths were compared with clinical variables.
Results. For a set number of implants, the relationship between calculated cantilever length and anterior-posterior spread was linear. The sum of length on both sides versus prosthesis length between the most distal implants was linear, regardless of the number of implants. Predicted satisfaction was defined as calculated length greater than the clinicians' optimal length. Satisfaction rates were 100%, 56%, 33%, 8%, and 0% for cases supported by 8 and 7, 6, 5, 4, and 3 implants (44% overall), respectively. Ninety-eight percent of cases with anterior-posterior spreads greater than 11.1 mm were satisfied.
Conclusion. Within the limitations of the model, predicted complications of the gold screw joint may be reduced if: (1) cantilever length is less than calculated from linear equations, and (2) anterior-posterior spread is greater than 11.1 mm. (J Prosthet Dent 2000;83:332-43.)

CLINICAL IMPLICATIONS

The frequency of gold screw loosening may be reduced if the prosthesis has a shorter cantilever. Implant distributions with anterior-posterior spreads greater than 11.1 mm may allow a cantilever sufficient to provide satisfactory biomechanics, esthetics, phonetics, and function without gold screw biomechanical complications.

The success of osseointegrated implant-supported prostheses in restoring structure, function, phonetics, and esthetics is a function of biomechanics, surgery, and biomaterials. The biomechanics of the implant-supported prosthesis system is complex. The biomechanical response to applied loads is a function of the mechanical and geometric properties of bone, implants, components, super structure, teeth, and their interfacial properties. Examples include the number, size, and shape of implants; the location of implants in relation to bone and other anatomic structures; quality of bone; the occlusal plane and scheme; distribution of implants along the arch; choice of components and retention at the abutment and at the prosthesis level; and prosthesis design. Additional complications are due to the nature of the intraoral loading system itself, including force locations, magnitudes, directions, cyclic patterns, and boundary conditions. Sequential addition of the above individual biomechanical variables can cause geometric increases in applied loads to the system.[1] Furthermore, the mechanical and biologic responses may be intertwined.

Placement of implants with significant load-bearing capacity often may be restricted to the anterior portions of the arch. Thus, cantilevers distal to the most posterior implants are often required. The presence of load-bearing cantilevers increases the forces distributed to implants, possibly up to 2 or 3 times the applied load on a single implant, due to bending moments.[2-7] To avoid mechanical overloading, the length of the cantilevers used clinically is often lower than those deemed by the clinician to be optimal for restoration of structure, function, and esthetics. Hence, the effects of cantilever lengths (CLs) on occlusal forces distributed to implants have been extensively studied. Several clinical

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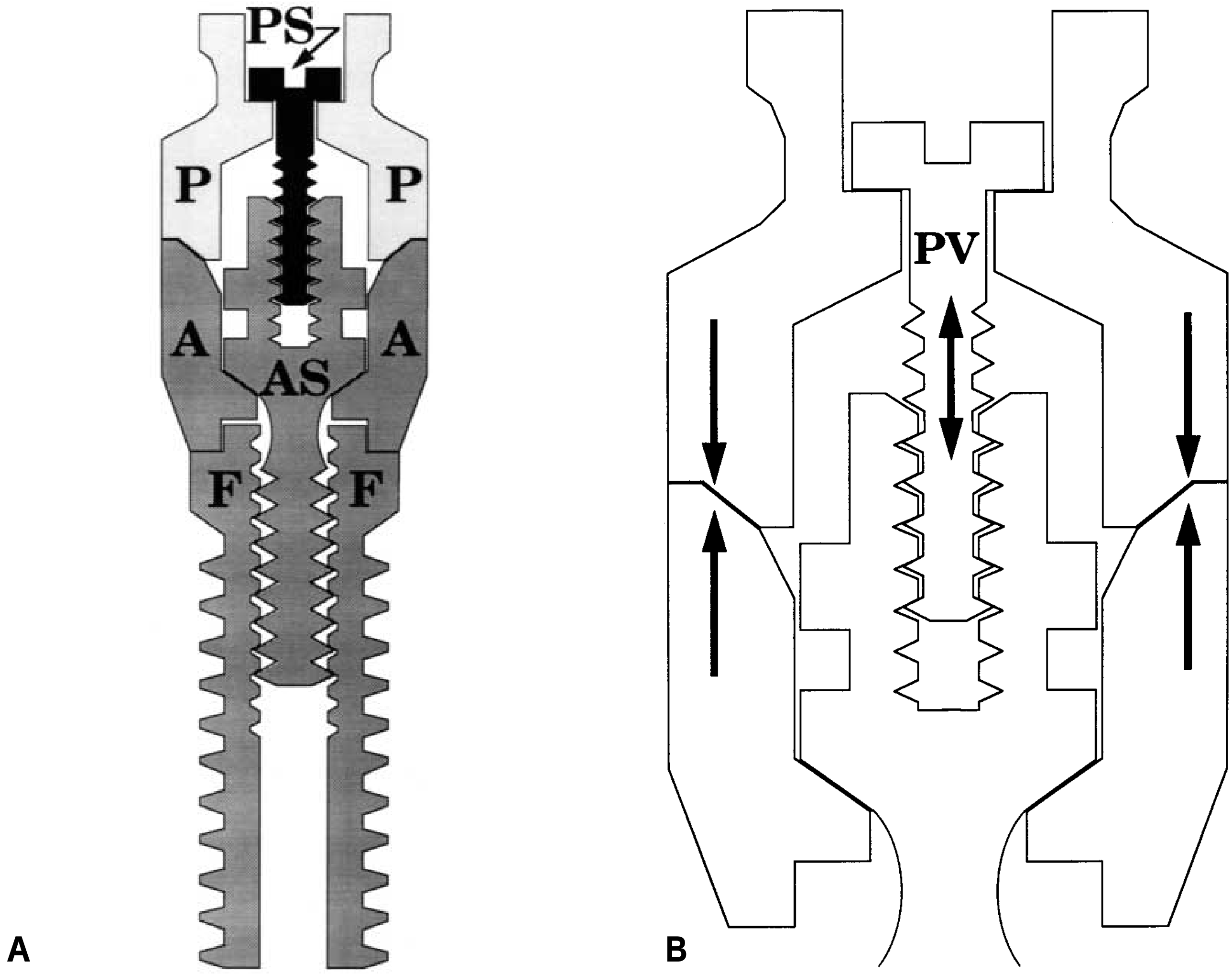

Fig. 1. A, Cross section of implant connected to abutment and prosthesis gold cylinder. *F*, implant, *A*, abutment, *P*, prosthesis gold cylinder, *AS*, abutment screw, and *PS,* prosthesis-retaining (gold screw). All theoretical calculations in study are based on this retentive system. **B,** Close-up view of prosthesis abutment interface cross section. When prosthesis is in place and no loads are applied to it, joints between prosthetic components (P-A, P-PS, and PS-AS) are held closed (compressive force *arrows* in P and A) because of preload (*PV*) in prosthesis retaining screw.

guidelines regarding the maximum cantilever lengths usable without causing biomechanical complications have been suggested.[4,7-15] The guidelines are often a function of the number and distribution of implants and the arch in which they are placed.

Biomechanical overloading may result in complications at the bone to implant interface, within the prosthesis and components, or at prosthetic joints.[16] Clinical studies have shown that biomechanical complications at the bone to implant interface, such as (1) implant fracture, (2) loss of osseointegration, and (3) bone fracture, are comparatively rare and usually appear after the complications at the prosthetic components and their joints.[1,16]

The final attachment of the prosthesis to an implant can be through screw retention, cementation, or a combination of both.[17,18] One advantage of screw retention is greater retrievability.[17,18] In the screw-retained Brånemark system, the prosthetic screw joint has been designed to be the “weak link” because its repair is less complicated than that of lower components or loss of osseointegration.[1,19] Complications include screw loosening, plastic deformation, or fracture. Loosening of the prosthesis-retaining screw occurs at the lowest loads when compared with all other structural complications.[1,19] Consequently, loosening of the prosthesis-retaining screw is the most frequent complication encountered with this design and is

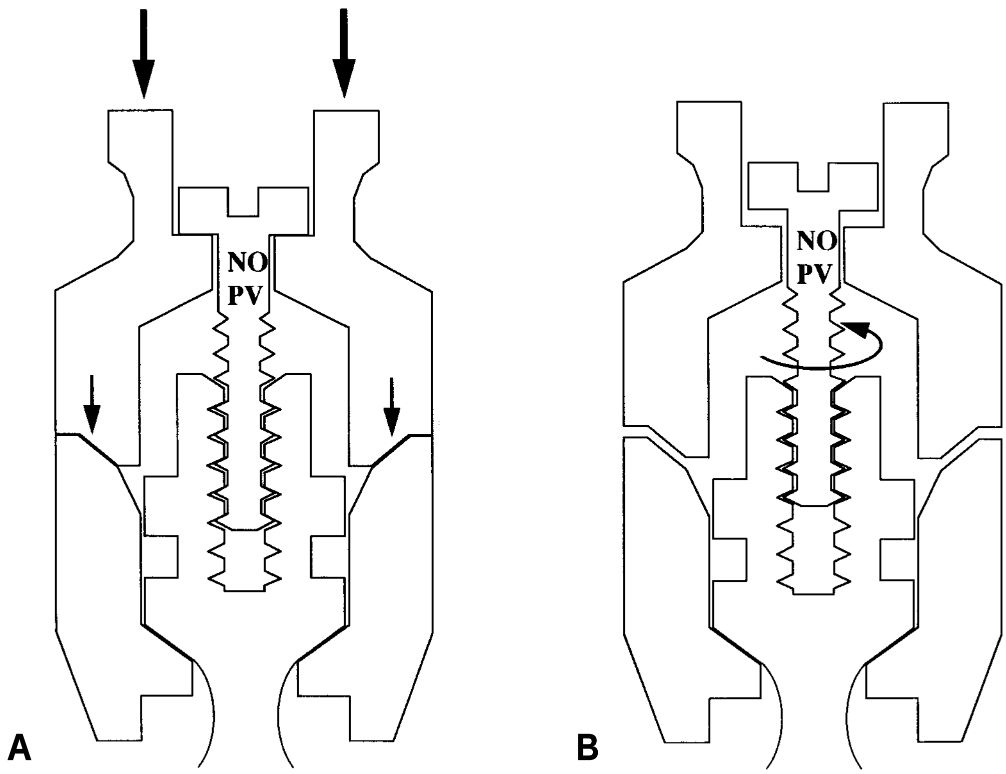

Fig. 2. A, Under compression, load is transferred to abutment through surface area of P-A joint (*small black arrows*); P and A elastically deform lessening clamping force at PS screw head–P interface. If load reaches or exceeds preload, prosthesis-retaining screw is not loaded with any preload. **B,** Possible result of excessive compressive loads (exceeding preload) is loosening of prosthesis retaining screw and opening of joints (P-A, P-PS, and PS-AS). Because screw is no longer held in place, rotational or shearing forces in horizontal plane, represented by *arrow*, may cause loosening.

believed to be the most common indicator for mechanical overload.[1,19] The percentage of prostheses-retaining screw loosening reported in the literature has ranged from 5% to 49%.[16,20-23]

The opening of a joint at the prosthesis-retaining screw–prosthesis interface can be due to either excessive compressive or excessive tensile forces.[16,18,24-26] During compression, joint opening can be caused by screw loosening if the preload is exceeded. The preload of a screw or bolt is the tension developed in the screw because of the applied torquing forces during screw tightening (Fig. 1).[27] Because it is the tension in the screw that holds the clamped pieces together, when a compressive force of equal or greater magnitude is applied, screw loosening may occur (Fig. 2).[27] Tensile forces, due to bending moments, can cause a joint opening due to plastic deformation or fatigue of the interface components such as the gold screw (Fig. 3).[24,28] A joint opening will reduce or eliminate the capacity of this implant to carry tensile loads, possibly causing greater forces on the remaining implants.[6] Tensile loads can lead to fatigue failure of the gold screw because of the cyclic nature of occlusal loads.[24,28] Also, if there is an opening, occlusal loads may cause impact loading of this possibly strain-hardened (therefore more brittle) screw, leading to screw fracture.

The purpose of this study is to investigate the effects of clinical variables on maximum cantilevers usable without mechanical overloading. To this end, the maximum permissible cantilever is defined as one in which the force on any implant does not exceed either the gold screw preload in compression, or cause fatigue failure in tension. Investigation into possible relationships between calculated cantilever length clinical variables included number and distribution of implants, arch placement, and the clinically optimal cantilevers. The distribution of the applied vertical load to implants on 55 clinical cases was calculated theoretically using the Skalak model.

MATERIAL AND METHOD

There were 4 major steps to this investigation: (1) obtain casts of clinical cases, (2) input arch geometry and implant locations into the computer, (3) calculate

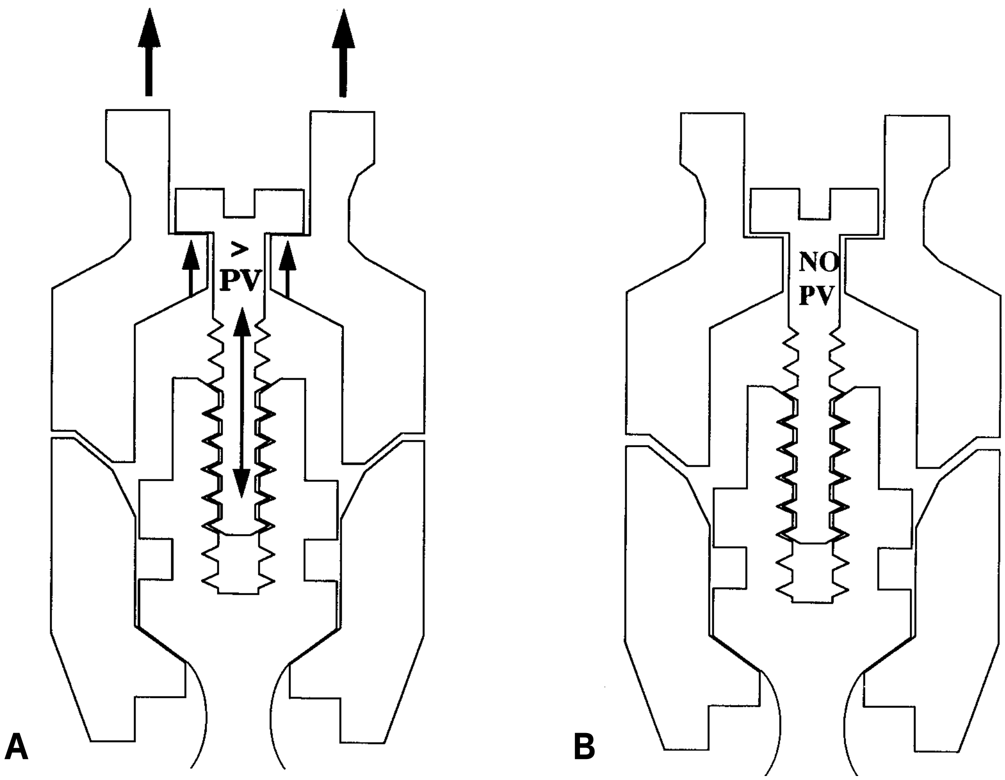

Fig. 3. A, Prosthesis-retaining screw joint may be subjected to tensile loads. Tensile loads may lead to plastic deformation, fatigue, or fracture of prosthesis-retaining screw. **B,** Excessive tensile loads may cause opening of joint.

the maximum permissible CL through the Skalak model, assuming mechanical overload to occur when the prosthesis-retaining screw–prosthesis joint is compromised, and (4) analysis of these calculated CL versus number of implants, anterior-posterior (AP) spread, arch placed, clinicians' optimal CL, and clinically reported biomechanical concerns.

Fifty-five clinical cases were analyzed (Tables I and II), of which 44 were gathered from 7 private offices maintained by prosthodontists or general practitioners in the New York and New Jersey areas. Eleven cases were obtained from the graduate prosthodontics clinic at Columbia University. The clinical information required was the (1) number and geometrical arrangement used for a fixed complete implant-supported prosthesis, (2) positions of the most distal locations of the prosthesis desired by the restorative clinician for optimal function, esthetics and phonetics (not necessarily those used), and (3) clinical outcome for this implant distribution, if available.

Therefore, the selection criteria for inclusion in this study include that each of the following was available:

1. a treatment plan for a fully implant supported arch prosthesis;
2. a cast of the edentulous ridge with abutment analogues in place;
3. information about the opposing dentition;
4. length of the extension of the prosthesis posteriorly to the most distal implants, desired by the patient and restorative dentist for optimum function, esthetics, and phonetics (this length was used as the "optimal" CL for the predicted satisfaction of each case in this study);
5. the actual CL on each side of the arch, if the case was already restored; and
6. documentation of any complication if the prosthesis had been under function for any period, such as screw loosening or failure. To maximize the number of cases available for this study no other selection criteria were used.

Table I. Number of cases by number of implants and by arch

Implant no.	Case no.	Maxillary cases	Mandibular cases
3	2	1	1
4	13	4	9
5	15	5	10
6	18	11	7
7	2	1	1
8	5	3	2
Total	55	25	30

Table II. Cases categorized by number of implants

Implant no.	Case no.	AP Mean ± ½ SE	Average CL Mean ± ½ SE	CL/AP Mean ± ½ SE
3	2	5.68 ± 0.05	5.81 ± 0.18	1.02 ± 0.04
4	13	7.03 ± 0.53	8.64 ± 0.54	1.30 ± 0.04
5	15	10.26 ± 0.65	15.43 ± 0.87	1.56 ± 0.03
6	18	12.79 ± 0.57	23.40 ± 0.98	1.87 ± 0.02
7	2	13.83 ± 1.42	35.85 ± 0.45	2.69 ± 0.25
8	5	20.31 ± 0.98	52.31 ± 2.62	2.57 ± 0.01
Total	55	11.20 ± 0.27	20.18 ± 0.96	1.71 ± 0.03

Mean values and SE of the anterior posterior (*AP*) spread, predicted cantilever length (*CL*), and CL/AP averaged on both sides of the arch. Note the increase of AP, CL, and CL/AP with number of implants.

All cases were grouped and analyzed by arch shape (square, tapered, and round) and by arch size (small, medium, and large). The cases were then categorized according to the symmetry of the placement of the implants along the arch as symmetrical, slightly asymmetrical, and asymmetrical. Implant positions and arch geometries were determined from photographs of the casts perpendicular to the occlusal plane. The occlusal plane was chosen because, in this study, the occlusal load is always applied to the prosthesis in a direction vertical to the occlusal plane. Cast bases were trimmed parallel to the occlusal plane at a constant base to plane height to reduce possible geometric artifacts due to out-of-plane rotations. Coordinates of implant and prosthesis locations were then digitized from photographic prints (Sigma Scan, Jandel Scientific, San Rafael, Calif.).

Distribution of an applied vertical force to implant-bone connections was calculated via the Skalak theoretical model. Input included the magnitude of the applied force and the coordinates of the applied force and implant positions, P, (x_r,y_r), and (x_{pr}, y_{pr}). The coordinates were translated so the origin was placed at the centroid of the implants, (x_c,y_c), $x_c = (\Sigma x_{ir})/N$, $y_c = (\Sigma y_{ir})/N$, $x_i = x_{ir} - x_c$, $y_i = y_{ir} - y_c$, $x_p = x_{pr} - x_c$, $y_p = y_{pr} - y_c$, where N is the number of implants, (x_i,y_i) and (x_p,y_p) are the centroidal coordinates of the implants and load positions. The force distributed to an implant, F_i, was calculated from the following formula:

$$F_i = P / N + P(A \cdot x_i + B \cdot y_i)$$

$$A = \frac{(I_{xy} \cdot y_p - I_{xx} \cdot x_p)}{(I_{xy}^2 - I_{xx} \cdot I_{yy})}$$

$$B = \frac{(I_{xy} \cdot x_p - I_{yy} \cdot y_p)}{(I_{xy}^2 - I_{xx} \cdot I_{yy})}$$

$$I_{xx} = \Sigma y_i^2;\ I_{yy} = \Sigma x_i^2;\ I_{xx} = \Sigma x_i y_i$$

For each case the load position was varied for at least 500 points along the prosthesis (range 557-1024), depending on the number of points automatically digitized while tracing the arch. The occlusal load input was 143 N.[29-31] Failure of the prosthesis was assumed to occur when forces on any one implant, as calculated through the Skalak model, were either greater than 200 N in compression,[25,26] or greater than 250 N in tension.[24] Predicted satisfaction was assumed to occur if the prosthesis could be extended posterior to the optimal cantilever lengths without exceeding the maximum permissible CL predicted by the model.

One-way analysis of variance (ANOVA) was used for all statistical comparisons. Statistical comparisons include tests for variations in AP spread (Fig. 4), CL, and CL / AP ratio with arch, arch form, arch symmetry, and number of implants. CL versus AP plots were constructed and linear equations were fit to the plotted data through a least squares minimization. Because there were only 2 cases of 3 implants and 2 cases of 7 implants, the constants in the linear equations for these cases were obtained through extrapolation and interpolation of the curves of the cases with 4, 5, 6, and 8 implants cases, respectively. Similarly, straight-line fits were performed for CL versus the prosthesis length between the most distal implants. A plot of the difference between the calculated maximum permissible CL and the optimal CL versus AP was also constructed. Finally, the optimal CL was added to the AP to determine whether, in general, the clinicians had a tendency to obtain a constant prosthesis length in all clinical cases.

RESULTS

When an occlusal load was applied on a cantilever area, the most distal implants carried compressive loads and the implants closest to the midline carried tensile loads (Fig. 5). Excessive loads always occurred when the occlusal load was applied to the cantilever areas. Compressive forces carried by the most distal implants can reach values of 2 or even 3 times the value of the applied occlusal load. Compressive forces of 200 N always appeared before tensile forces of 250 N. Therefore, tensile forces did not affect the calculated CL, AP

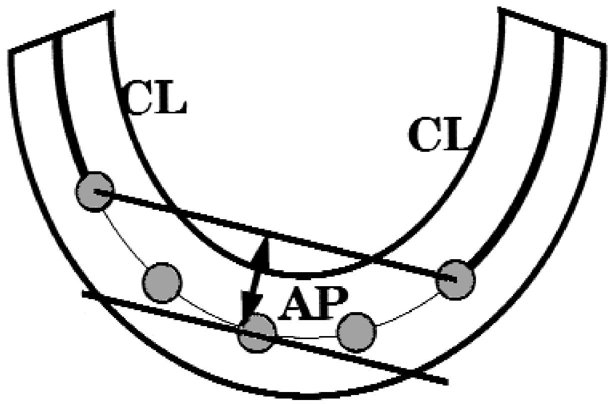

Fig. 4. Schematic representation of cantilever length (*CL*) and anterior posterior spread (*AP*) in hypothetical 5 implant cases. CL in implant prosthesis is length of overstructure projecting distally from most distal implants. AP in implant prosthesis is distance between line connecting 2 most distal implants and its parallel line passing through center of implant most distal to that line.

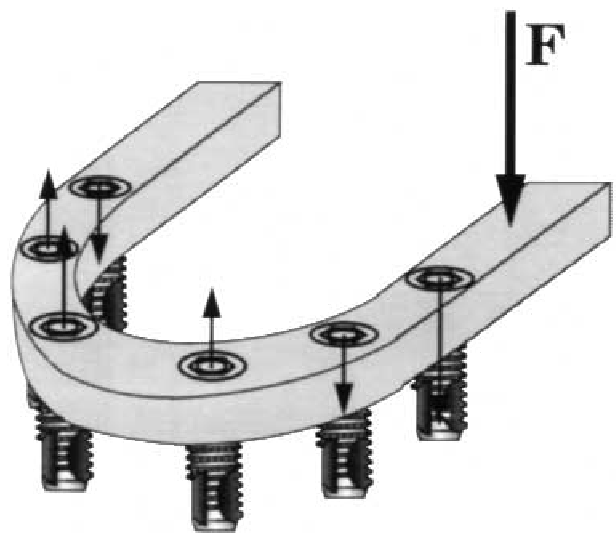

Fig. 5. Three-dimensional schematic of arch form prosthesis supported by 6 implants. Occlusal load (*F*) applied perpendicularly to occlusal plane on different points along prosthesis; distribution of that load to implants was calculated from theory. Load magnitudes and directions represented by *arrows*. Here occlusal load was applied along cantilever of one side of prosthesis and distributed load was calculated with Skalak model.

spread, or CL to AP spread ratio. For loads applied between the most distal implants, the highest implant force was 158 N in compression and 86 N in tension (3 implant cases).

The 55 cases were almost equally divided between the mandible and maxilla (Table I). The average AP and calculated CL of the maxilla are larger than the mandible, 13.7 mm and 24.8 mm versus 9.1 mm and 16.5 mm, respectively. CL / AP ratios of the 2 arches were statistically equivalent; 84% of the cases were treatment planned for implant-supported restorations on 4, 5 and 6 implants (Table II). No clinically significant differences that depended on the arch shape or size were found in the distribution of the forces along the arch, or the values of permissible CL and CL / AP ratio. The only exception was the large size cases that had a statistically larger CL than the medium- and small-size cases. Mandibular cases presented significantly higher asymmetry than the maxillary cases.

The values of AP increased with an increase in the number of implants. Predicted CL increased with both AP and number of implants (Table II and Fig. 6, *A* through *C*). For a particular number of implants, the relationship between calculated CL and AP was linear. Equations were CL = 0.50 AP + 3.40, 0.98 AP + 1.77, 1.33 AP + 1.80, 1.65 AP + 2.30, 2.18 AP + 0.02, 2.65 AP – 1.56 for 3, 4, 5, 6, 7, and 8 implant cases. The slope of these curves versus the number of implants was also linear (0.42 N-0.76) (Fig. 6, *D*). The data for CL versus actual prosthesis length between the most distal implants data was also linear. The shorter cantilever of each case caused larger scatter. This scatter was greatly reduced by plotting the sum of the cantilevers on both sides of the arch $CL_{left} + CL_{right}$ = 0.58 prosthesis length of –12.60 (Fig. 6, *E*). The mean value of the sum of the clinically optimal CL and the AP for all cases was 31.1 mm (standard error 0.7 mm).

Predicted satisfaction varied greatly and depended on the number of implants and the AP. The rate varied from 100% for cases with 7 or 8 implants, to 56% for cases with 6 implants, to 33% for cases with 5 implants, to 8% for cases with 4 implants; the rate was 0% for cases with 3 implants (Fig. 7). A total of 98% of the cases with AP spread higher than 11.1 mm, regardless of the number of implants, resulted in calculated cantilevers greater than the optimal length (Fig. 7).

Finally, in 6 of the 55 cases, for which some biomechanical complications / concerns were documented, the clinical outcome matched that predicted by the model. A 4-implant case, AP = 4.6 mm, exhibited repeated screw loosening and several fractures at the 2 most distal implants. Clinical CL exceeded the CL calculated through the model. Four more implants were placed increasing the AP to 220 mm. The new predicted CL was more than twice that used clinically and no loosening or fracture of prosthesis-retaining screws occurred. In another case, 5 implants were placed, but only 4 were used because of the close proximity of 2 adjacent nondistal implants. The model predicted over-

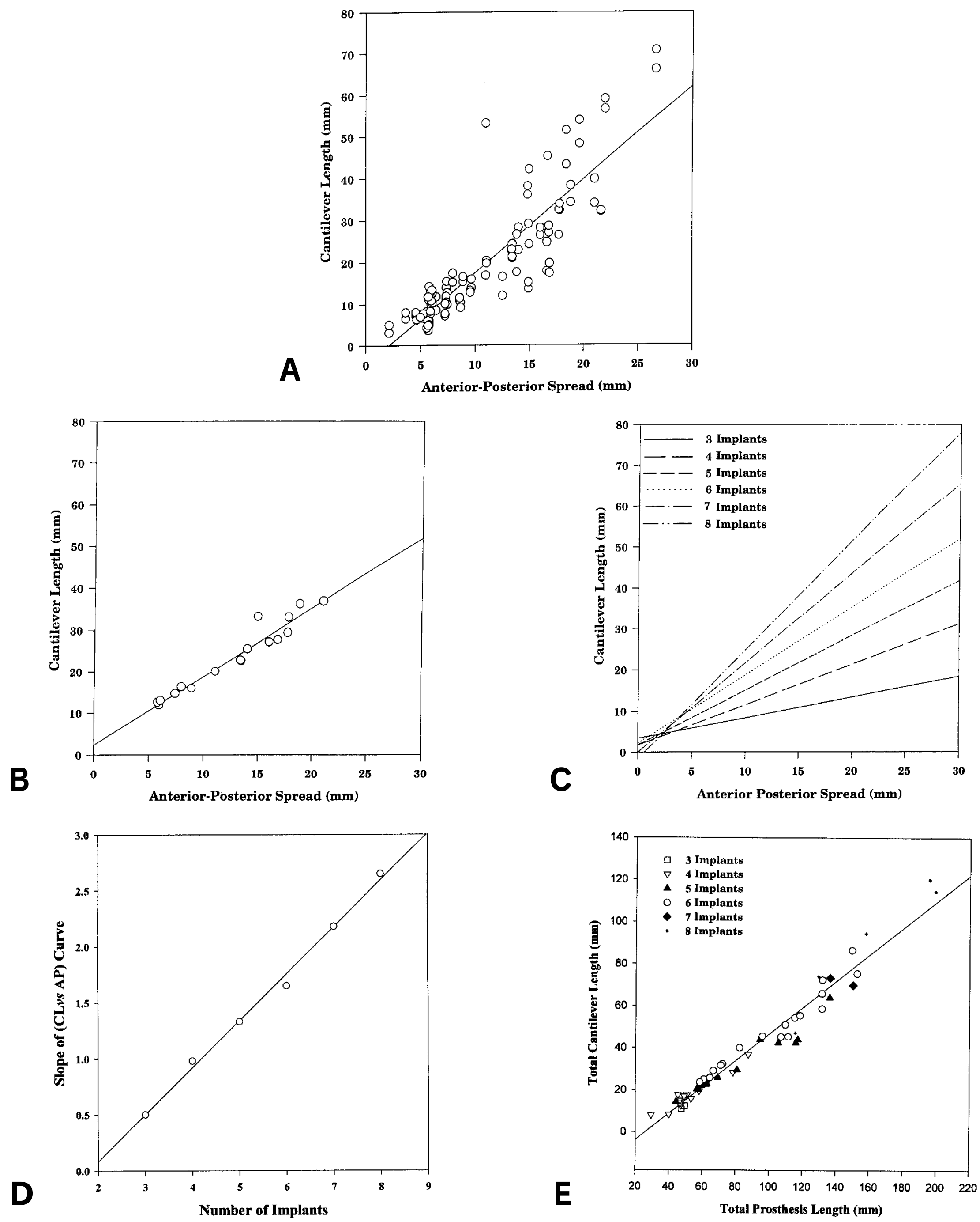

Fig. 6. A, Theoretically predicted CLs calculated through Skalak model versus actual AP spread for all 55 clinical cases. Cantilever lengths represent maximum cantilever possible without compromising biomechanical integrity of prosthesis-retaining screw joint. Predicted CLs increase with increasing AP spreads. Although there is general linear relation between these values, scatter is considerable. **B,** Theoretically predicted CLs calculated through Skalak model versus actual AP spread for all clinical cases supported by 6 implants. There is much less scatter in linear relationships between predicted CL and AP spread when cases with same number of implants are plotted separately. **C,** Linear relationships between theoretically predicted CLs versus actual AP spread by number of implants. As with **B**, straight lines for a fixed number of implants were fit to data. Parameters of linear equations are presented in text. Predicted cantilever increases not only with anterior posterior spread but also with number of implants. **D,** Linear relationship between slopes of lines in **C** versus number of implants. **E,** Sum of theoretically predicted CLs, as calculated through Skalak model, for both sides of arch versus actual prosthesis length between most distal implants for all 55 clinical cases. Plots of all individual CLs present more scatter.

loading with 4 implants but not with 5 implants. After similar overloading complications with only 4 implants, the prosthesis was modified to include the fifth implant and no screw loosening occurred. In another case, 5 implants were placed with an AP of only 3.6 mm. The model again matched clinical results because overloading was predicted and the prosthesis was shortened to an unsatisfactory length because of overloading. Two additional implants were placed in the extreme posterior region.

In another case, 8 implants were initially placed. Because of an unfavorable inclination of the 2 distal implants on the same side, implants were submerged under soft tissue. The clinician was concerned that, despite the use of 6 implants, overloading might occur as a result of the long pontic area. The model predicted no overloading, and follow-ups over the course of 2 years showed no loosening or fracture of the screws.

DISCUSSION

The relationships found in this study between theoretically calculated maximum CL and clinical distributions of implants include (1) maximum permissible CL versus AP was linear for a set number of implants, (2) predicted CL exceeded the clinicians' optimal CL for cases with AP greater than 11.1, and (3) the sum of the cantilevers on both sides of the arch versus the length of the prosthesis between distal implants was described by a single linear relationship. The above relationships were derived using parameters that closely mimic clinical reality; these parameters included an applied intraoral force in the in vivo range for implant-supported prosthetics, the lowest measured values for biomechanical complications (compromise of the prosthesis-retaining screw joint), and arch forms, number and positioning of implants from actual clinical cases being restored. The Skalak model was used to calculate the distribution of the applied load to implants.

The results of our study correlate well with CL clinical guidelines in the literature. Maximum acceptable CL guidelines include those based on lengths in millimeters or tooth size, implant distributions in AP or offset, and length of the prosthesis between distal implants. Sizes of suggested maximum CL vary with the number of implants and whether the implants are in the maxilla or mandible. Examples include the size of 1 or 2 teeth[8,10,19] and various CLs from 0 to 20 mm,[3,11-14] with lengths increasing with number of implants and placement in the mandible. Maximum CLs from distribution relationships include use of AP greater than 10 mm,[11] twice,[3] or 1.5 the AP,[14,15] and linear relations for a set number of implants,[7] offset of 1 or 2 abutments for the partially edentulous patient.[1] Determination of cantilever lengths from prosthesis lengths included prosthesis length minus 20 mm[3] and other linear functions.[7]

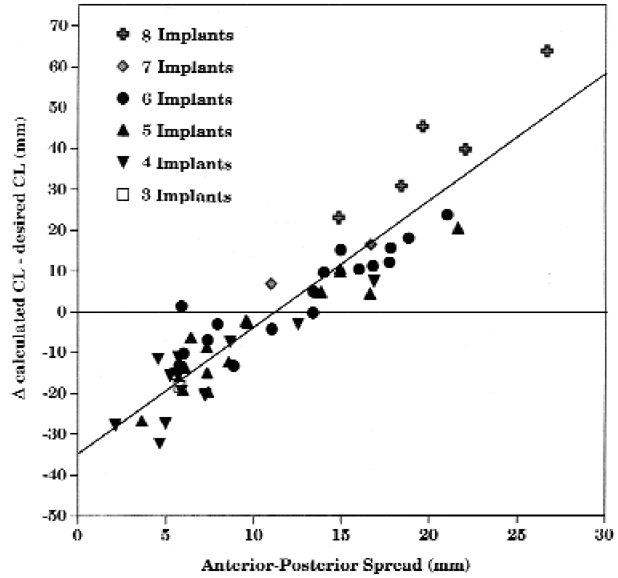

Fig. 7. Plots of difference between theoretically predicted CLs and CLs thought to be optimal by restoring clinicians for full restoration of structure, function, esthetics, and phonetics of 55 clinical cases. For AP spreads greater than 11.1 maximum cantilever predicted by model is greater than that thought to be optimal by restoring clinician.

Adherence to the maximum CL calculated by these guidelines may sometimes lead to prostheses that are biomechanically sound but unacceptable to the patients because of esthetics, phonetics, or function.

The distribution of implants, with respect to interimplant distance, implant offset, and AP spread affects the distribution of applied developed forces and moments to single implants.[1,33,34] Increasing implant offset decreases the loads and bending moments distributed for both fully and partially edentulous patients.[1,19] Thus, the maximum cantilever can be increased by increasing the AP (offset).

Our results demonstrated that, for clinical implant distributions, the relationship between CL and AP spread was linear but was a function of the number of implants. The average y intercept was 1.5 mm, which is probably clinically insignificant in this study and therefore can be ignored. The resultant ratios were 1.33 and 1.65 (avg. 1.49) for 5 and 6 implants, which is close to the suggested ratio of 1.5 for 5 and 6 implants. As predicted by the presented model, the use of 1.5 would not cause overloading for 6 implants but might for 5 implants. The use of a ratio of 2 is only appropriate for 7 and 8 implant cases. The results of this study are well within the range of those previously suggested.

A minimum AP of 10 mm has been suggested to provide a biomechanically acceptable CL.[11] This guideline correlates well with our finding that cases with greater than 11.1 mm provided cantilevers that equal or exceed those that are clinically optimal for full restoration of structure, function, and esthetics. In cases with larger AP, the most posterior implant is usually placed in a more distal position in the arch, which decreases the CL required to satisfy the functional and esthetic needs of the patient. Because as the AP not only increases the ability to cantilever increases, but also the clinical need to CL decreases as well, it is not surprising that the sum of the AP and the clinically optimal CL varies little from case to case (mean value = 31.1 mm; SE = 0.7). Hence, the most distal point clinically desired for a prosthesis to extend is relatively constant. As a result, the rate of clinical satisfaction (clinicians' optimal CL < calculated CL) of cases regardless of the number of implants increases drastically as the AP increases.

Results also were compared for cantilever guidelines that state specific lengths. No complications occurred under lengths of 36 mm for 8 implant cases. The model predicted that 20 mm was too long for most cases; complications were predicted for one of the two 7-implant cases, 33% of 6 implants, and 93% of 5 implants. The 15-mm implants fared slightly better with predicted complications for 6, 5, and 4 implant cases (22%, 53%, and 92%, respectively). For 12-mm cantilevers overloading was predicted for 6, 5, and 4 implant cases (11%, 40%, and 85%, respectively). With 10-mm CL, complications were predicted for 0%, 20%, and 85% for 6, 5, and 4 implant cases. To obtain a 10-mm cantilever length through the model, AP spreads of 6 and 10 mm were required for 5 and 4 implant cases. Therefore, the model best matched the midrange guideline of 10-mm cantilever for cases with greater than 4 implants. Again the suggested 10-mm AP spread guideline seemed to be appropriate.

The sum CL on both sides of the arch versus length of the prosthesis between implants was linear, regardless of the number of implants. Linear relationships have been found previously.[3,7] Cantilever estimates in our study were lower than the prosthesis length, minus 20 mm, found previously,[3] due to the lower more clinically based biomechanical compromise loads and clinical distributions and arch geometries used. Both relationships can only be used for symmetrical implant distributions. Asymmetrical cases can cause a significant difference in CL on opposite sides of the arch.

Increasing the number of implants also increased the satisfaction rate (Fig. 7). The cause of this proportional relationship is 2-fold: (1) increasing the number of implants decreased the predicted force distributed to the individual implants ($1/N$ term in F_i equation) enabling longer CL, and (2) clinically, there was an increase in AP with an increase in the number of implants (Table II). For example, all cases supported by 7 and 8 implants were satisfied 100%. For 5 or 6 implant cases, only 33% and 56%, respectively, predicted to be safely restored to the desired length without concern. In these cases, the spacing and the positioning especially of the most distal of the implants become significant for the ability to extend the prosthesis posteriorly. These cases are often regarded as marginal and require more frequent recall and follow-up to secure biomechanical soundness of the prosthesis. Cases with 3 and 4 implants present almost no predicted satisfaction (0% and 8%, respectively). Cases with 3 or 4 implants may frequently be restored with prostheses that are not supported solely by the implants, but by removable prostheses sharing the load between the implants and edentulous ridges. Previously, percentage changes in AP tended to have a greater effect than equivalent changes in number of implants.[15]

Overall predicted satisfaction rate was 44%. Although percentage of satisfaction may seem low, it is assumed that failure occurred when the preload was exceeded, which may indicate screw loosening. The percentage of screw loosening observed clinically was also high, with reported values of up to 49%.[16,20-23] For all cases where the clinical outcome with respect to overload was known, the clinical result matched those predicted by the model. Because of the low number of cases in which clinical outcomes are known (6 of 55 cases), the above comparison is anecdotal. Longitudinal outcome assessment would be required to state whether a correlation between the model prediction and clinical outcome exists. Regardless, the preliminary correlation is suggestive.

Trends were found with the positioning and number of implants placed versus the arch restored. The mean value of AP for all the mandibular cases was 9.1 mm and for all the maxillary cases 13.7 mm. When the same number of implants was used, the implants were placed closer to the midline and to each other in the mandible, whereas in the maxilla they are more widely distributed along the arch. The above phenomenon may be due to the different anatomy of the mandible and the maxilla. The mental foramina and the intraosseous route of the mandibular nerve often restrict the available bone sites in the mandible more than the maxillary sinuses do in the maxilla. Also, practitioners have been encouraged to place implants widely in the maxilla because of lower maxillary bone density and the small proportion of cancellous bone, which may lead to lower osseointegration maxillary rates.

Placement of the implants in the mandibular arch was statistically more asymmetrical to the midline than in the maxillary arch. Higher esthetic demands in the maxillary arch dictate a more precise placement of implants at sites corresponding to teeth placement, forcing practitioners to place implants in the maxilla in sites that present more symmetry. In the mandible, the placement of the implants in the anterior area is often not as critical for the esthetic success of a case, so placement of the implants usually does not follow symmetry to the midline.

The number of cases categorized by number of implants revealed that the most common were those with 4, 5, and 6 implants (84% of the total number of cases, Table I). Only 4% of cases were restored with 3 implants (cases of maxillectomy and hemimandibulectomy) because 3 implants are regarded as insufficient to support a complete arch totally with implant-supported prosthesis. Only 4% and 9% cases were restored with 7 and 8 implants, respectively. Despite the increasing concern of biomechanical failures and the emphasis on "over-engineering" of implant prostheses, the use of 4, 5, and 6 implants for restoration of edentulous arches with implant-supported restorations appears to be common. The information from our study regarding the number and distribution of implants may contain a geographical bias because all cases were collected from the New York metropolitan area.

Limitations of the study are due to the assumptions in the model used to calculate the distribution of applied load to the implants and the values for applied and joint compromise loads, and have previously been described.[7] Distribution of applied occlusal load to implants was calculated through the Skalak model. In the Skalak model, the connection between bone and prosthesis is elastic (deflection is proportional to the load), whereas the bone and prosthesis are relatively stiff.[2] Input to the model included the position, direction, and magnitude of an applied load and the number and distribution of implants.

All input and output are approximately at the abutment level (gold screw joint level). Analysis of forces at the abutment level may be appropriate assuming the gold screw joint as the weak link. Also, this gold screw joint is the most flexible part of the bone / implant / prosthesis system,[28] so the Skalak assumption of elastic connections is appropriate. Skalak calculations correlated well with in vitro experimental measurements of axial abutment forces due to the distribution of applied load on a gold superstructure supported by 5 implants in cadaver bone.[6] The correlation was even better than a more complex model.[6] Thus, Skalak's method was recommended for predicting loads transmitted to the abutments.[6] This close correlation was suggested for implants in bone in earlier experimental study with nonrigid superstructures supported by 6 implants in an aluminum block[35]; later experiments confirm this close correlation.[36] In another model, when typical values were used, the distributed forces differed only slightly from those calculated through the Skalak model.[37] Again the Skalak model was claimed to be accurate for estimating the vertical force distribution.[37] Although a direct comparison cannot be made (because the exact geometry is not known, artifactual asymmetry exists in their experiments, and the sum of the forces does not equal zero), Skalak results calculated in our study correlated well with another in vitro connections force distribution study, especially for their supported case.[51]

The Skalak model can be easily implemented with a small personal computer or programmable calculator. Therefore, although the many biomechanical factors are not included in this analysis—bone quality, quantity, and deformation; implant size, shape, and relative angulations; prosthesis design; and occlusal scheme,[1] the Skalak model may be appropriate to attempt to separate out the effects of simple variables such as number and distribution of implants and CL on forces at the abutment level. Other methods, such as direct measures or finite element and experimental analysis, also have assumptions, limitations, and errors.

The applied 143 N load was chosen because it is a mean maximal load (range 42 to 412).[29-31] Lower or higher applied loads would enable longer or shorter cantilevers, respectively. A 143 N vertical force is also in the range of other studies[38,39]; therefore it is reasonable. Loads distributed to implants that may compromise the prosthesis retaining joint are based on the Brånemark system with a gold flat-head–designed prosthesis-retaining screw. Compressive loads greater than the preload of 200 N were assumed to lead to gold screw loosening. A load of 200 N was the higher preload developed with the cast-to as opposed to the as-received cylinders.[26] Clinically obtained preload is a function of many variables, including torque applied, method of applying torque, joint materials, design, lubrication, and mating surfaces.[6,17,18,25,26,28,40-49] Although higher preloads can be obtained and antirotational devices can be used, 200 N is still reasonable as a result of the complex interaction of the above variables.

Compromise due to tensile loads was assumed to be 250 N in tension, which represents a tensile force on the joint where no fatigue effects are expected.[24] Fatigue loading was considered because occlusal loading is cyclic, the majority of metallurgical failures occur in fatigue, and fatigue events occur at loads much lower than the yield or ultimate tensile stress. Becasue tensile forces affect fatigue, 250 N is a safe magnitude for tensile loads on the gold screw joint. Other values, which could have been used, are higher than 250 N (50% or 65% of the ultimate tensile strength,[50] which is 300 and 390 N). Higher tensile compromise forces would not affect the results in this study because 200 N in compression occurred at lower cantilever lengths than 250 N in tension.

CONCLUSIONS

Within the limitations of this study, the following conclusions were drawn:

1. In 98% of all clinical cases studied with an AP greater than 11.1 mm, the maximum CL calculated through the model was greater than the CL desired by the clinician restoring that case.

2. For a set number of implants the calculated maximum permissible cantilever lengths as calculated through the model varied linearly with AP.

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Noteworthy Abstracts of the Current Literature

Multicenter retrospective analysis of the ITI implant system used for single-tooth replacements: Results of loading for 2 or more years

Levine RA, Clem DS III, Wilson TG Jr, Higginbottom F, Solnit G. *Int J Oral Maxillofac Implants 1999;14:516-20.*

Purpose. Replacement of missing single teeth by implant-supported crowns is a valuable treatment option because it provides all the benefits of dental restorations without the risk of recurrent caries. A preliminary report of single ITI implant-supported crowns showed high levels of patient satisfaction and implant survival. This article describes a multicenter retrospective study of single implant-supported crowns that have been in place at least 2 years.

Material and methods. Endosseous implants were placed and restored in private practices within the United States. Complications, such as implant failure, peri-implantitis, implant fracture, prosthesis loosening, abutment loosening, and prosthesis dislodgment were recorded. All implants were in place and followed for at least 2 years.

Results. A total of 174 ITI implants were placed in 129 patients with 110 patients available for followup. Reasons for dropout are described as patient death (n = 3), connection of an implant to a natural tooth (n = 2), no contact with patient (n = 8), and patient moved from the area (n = 6). Various designs of implants from the same manufacturer were used. Implant failure due to persistent peri-implantitis resulted in the loss of 4 implants after 6 months of function. Three implants failed because of implant fracture (mean time 40.3 months). All fractured implants were in the molar areas, demonstrating a molar fracture rate of 4% and an overall fracture rate of 1.9%. All fractured implants were 3.5-mm wide, hollow screw designs. Combination of failed and fractured implants results in an absolute implant survival rate of 95.5% for implants in function at least 2 years. Conical abutments loosened 4 times during the course of the study (5.3%). One crown was dislodged because of cement washout and 1 crown was remade as a result of “screw stripping.” Loosening of the octabutment occurred once, whereas loosening of the crown-retaining screws was seen in 11 restorations for a total of 22.2% of these restorations.

Summary. Implant survival, when single implants are used to support individual crowns, is acceptable at the rate of 95.5%. Single crowns cemented over conical abutments resulted in fewer complication than did crowns secured by screw retention. 11 References. *—SE Eckert*