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Effect of Engaging Abutment Position in Implant-Borne, Screw-Retained Three-Unit Fixed Cantilevered Prostheses

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Abstract

Purpose: To investigate the effects of internally connected engaging component position in screw-retained fixed cantilevered prostheses.
Materials and Methods: Twenty-one three-unit fixed dental prostheses (FDPs) were cast in high-palladium alloy in three groups. In group A, engaging components were incorporated into the units away from the cantilevered segment; proximal units received nonengaging components. In group B, these positions were reversed. Control specimens were fabricated using all nonengaging components. Specimens were attached to internally connected 3.5 (diameter) × 13 mm (length) implants, torqued to 32 Ncm, and embedded into epoxy resin. Specimens were tested in cyclic fatigue with a 2 Hz sine wave and 0.1 min/max load ratio. Load amplitude started at 1.8 N and increased by 1.8 N every 60 cycles until fracture. Log-rank statistic, ANOVA, Spearman's correlation, and LIFETEST procedures were used to evaluate level of statistical significance within the results.
Results: In the control group, the mean number of cycles to fracture was 31,205 ± 2639. Mean axial force at fracture was 932 ± 78 N. In group A, these numbers were 38,160 ± 4292 and 1138 ± 128 N, and in group B, 31,810 ± 3408 and 949 ± 101 N. Statistical significance levels for number of cycles to fracture were: Control versus group A, $p = 0.0117$, and groups A versus B, $p = 0.0156$ (statistically significant). Control versus group B, $p = 0.357$ (not statistically significant). Log-rank statistic for the survival curves is greater than would be expected by chance; there was a statistically significant difference between survival curves ($p = 0.012$). The location and mode of failure were noteworthy (always in the abutment screw).
Conclusions: The position of the engaging component had significant effects on the results. Within the limitations of this investigation, it can be concluded that using an engaging abutment in a screw-retained fixed cantilevered FDP provides a mechanical advantage, and engaging the implant furthest from the cantilever when designing a screw-retained cantilever FDP increased resistance to fracture of the distal abutment screw.

Due to the availability and location of viable osseous tissue, position, and configuration of adjacent teeth, their roots, and other vital structures, as well as poor surgical planning and execution, dental implants can commonly be presented to the restoring clinician when cantilevers are unavoidable (Figs 1, 2). Cantilevers have been studied extensively in the scientific literature, mostly pertaining to Branemark's original tissue-integrated prostheses for the fully edentulous mandible.[1] Empirical suggestions have been made,[2] finite element analyses,[3] in vitro[4-8] and in vivo[9-12] investigations have been conducted, and review papers have been published.[13-15]

Clinical reports and retrospective analyses have been published where complications such as implant fracture, prostheses loosening and decementation, and abutment screw damage have arisen in the presence of cantilevers, leading the authors to believe cantilever extensions need to be accounted for when considering prostheses design.[16,17] One author even suggests an alternative design to avoid cantilevered extensions altogether.[18]

Figure 1 A typical patient presentation with partial edentulism.

Figure 2 Treatment includes a cantilever extension.

Figure 3 Examples of anterior (A) and posterior (B) screw-retained fixed prostheses with one engaging component in each. The posterior prosthesis has one unit cantilevered.

Figure 4 Positioning analogs using cast-to cylinders and surveyor.

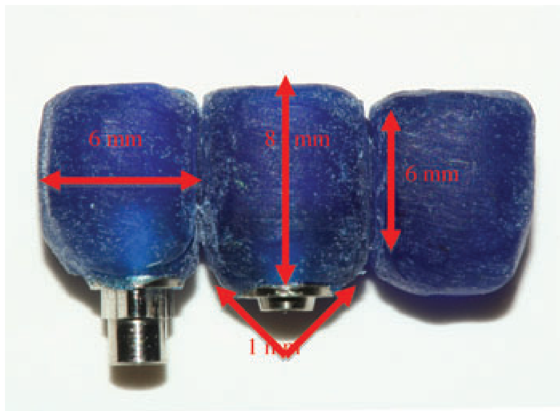

Figure 5 Master specimen.

Another component playing a crucial role in stability and longevity of implant-supported prostheses is the design of the implant-abutment interface. The body of evidence demonstrating and comparing precision and long-term performance of implant-abutment connection designs is significant.[19-25] Although there is conclusive scientific evidence that eliminating misfit and engaging the antirotational feature while applying adequate preload on the abutment screw significantly reduces screw loosening and other mechanical complications,[26-29] prior to the widespread use of internally indexed implant systems, screw loosening, and difficulty/inconvenience for the operator when seating restorations were commonplace occurrences in the practice of implant dentistry. This can be attributed in part to the relatively short lateral wall height of external indexing mechanisms (average 0.8 mm) when compared to an average of 2.4 mm of lateral wall engagement in internally indexed systems. Taking advantage of this mechanical component seems intuitive, but the very same fact that creates this biomechanical advantage also causes the seating of the multiple-unit splinted restorations to be difficult, if not impossible, in the presence of nonparallel implant fixtures when restoring them with splinted screw-retained prostheses.

In light of this information, unintentional as they may be, cantilevers must be accounted for during the restoration design

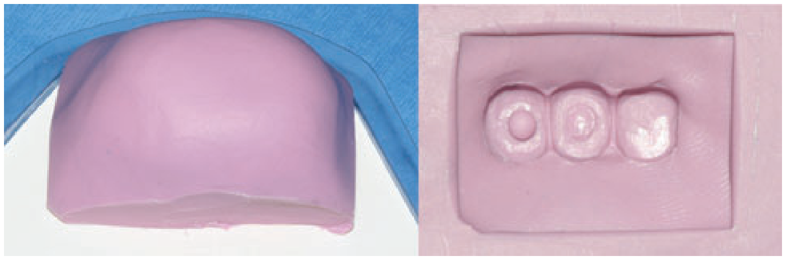

Figure 6 Investment of master specimen.

Figure 7 Waxing experimental specimens.

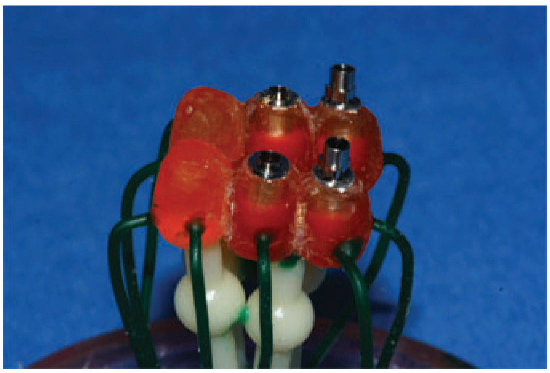

Figure 8 Spruing and investing specimens.

phase when replacing missing teeth with implant-supported fixed restorations. Many important decisions need to be made. Whether to splint these implants, or restore them as individual units, to retain the prostheses by way of screws or cement, whether to use custom or stock components, and designing the size and configuration of the occlusal table represent only a few of these important questions.

Mechanical failure of the implant/abutment/prosthesis complex can occur at different levels, each one requiring a separate set of interventions. Fracture of the implant fixture would require a surgical approach, whereas failure at the level of the abutment screw or prosthesis level can possibly be resolved by prosthodontic measures. Each time the design of the prosthesis is altered, the biomechanics of the whole complex is changed, and the weakest component of the equation may be moved.

Figure 9 (A) Control group. (B) Experimental group A. (C) Experimental group B.

Once the choice has been made to design a screw-retained connected prosthesis replacing multiple teeth, one question to be contemplated is whether to engage the antirotational feature and internal wall of one or more of the implants using an engaging abutment. Presently, there are no manufacturer

Figure 10 Specimen assembled in resin medium.

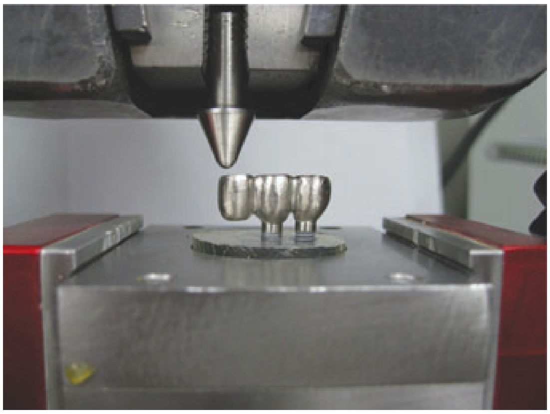

Figure 11 Specimen in servo-hydraulic test machine.

guidelines provided to guide this decision. A review of the literature yields no peer-reviewed evidence available with regard to engaging component selection in multiple connected units.

When restoring implants with connected screw-retained restorations, an institutional protocol uses at least one engaging component whenever possible. The use of one engaging abutment provides prosthetic convenience in positioning and

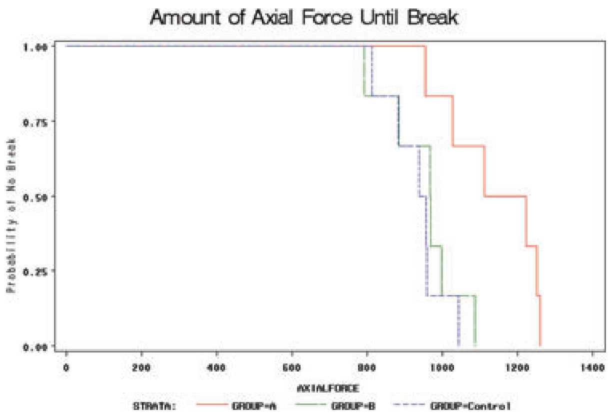

Figure 12 Survival curves for amount of axial force.

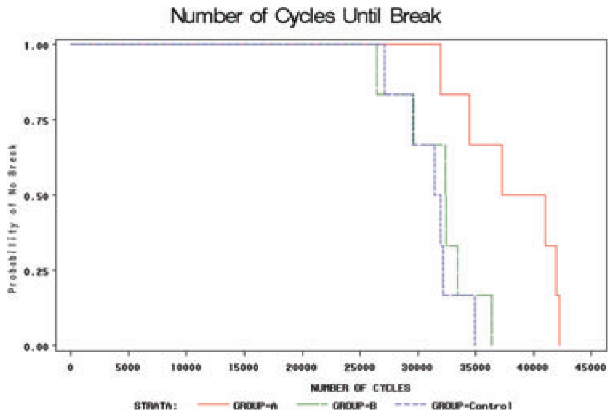

Figure 13 Survival curves for number of cycles.

seating restorations, and intuitively, can take some of the stress off the abutment screw and transfer it to the implant fixture, thus creating a different biomechanical complex. Due to the lack of direct relevant scientific data, this practice is based on anecdotal evidence and the clinical experience of educators and clinicians. Sometimes, treatment is completed with no engaging components; at other times, clinical factors such as implant position in the arch, implant length and diameter, the type of bone or bone graft in which the implant had been placed, implant depth, and amount of soft tissue collar around the platform is used to determine which implant to engage (Fig 3).

The purpose of this in vitro study was to investigate if engaging the antirotational component of one implant in an implant-supported, screw-retained three-unit fixed cantilevered prostheses yielded a mechanical advantage over using nonengaging components in all units, and to investigate the outcome of switching the position of the engaging component in the same prosthesis design.[30]

Materials and methods

Master model and specimen preparation

Two stainless steel, internally indexed narrow platform implant analogs (Replace Select, Nobel Biocare, Yorba Linda, CA) were placed in predetermined three-dimensional positions using a surveyor (Ney surveyor, Dentsply, York, PA) (Fig 4). Two gold/plastic cast-to abutments (Goldadapt, Nobel Biocare) were connected to the master model, and a master specimen representing a three-unit fixed cantilevered prosthesis was fabricated using pattern resin (Pi Ku Plast HP36, XPDent, Bre-Dent, Senden, Germany) (Fig 5). This master specimen was then invested on vinylpolysiloxane (VPS) laboratory putty (Lab Putty, Coldent-Whaledent, Basel, Switzerland) in two stages (Fig 6). Following polymerization, the investment was separated, the master specimen recovered, and the cast-to abutments connected to the master model in the desired configuration (Fig 7). Using the VPS investment, the master specimen was duplicated 21 times, sprued, and invested into phosphate-bonded investment (GC Fujivest II, GC Corp, Tokyo, Japan) as per manufacturer's specifications (Fig 8). These specimens were separated into three groups of seven specimens per group.

In the control group, both cast-to abutments were nonengaging (Fig 9A). In experimental group A, engaging components were incorporated into distal units (away from the cantilevered unit) (Fig 9B); proximal units (next to the cantilevers) received nonengaging components. In experimental group B, these positions were reversed (Fig 9C). Casting of all specimens were accomplished in high-palladium ceramic alloy (Advantage, Jensen Industries, Los Angeles, CA). Following devesting, specimens were finished to predetermined dimensions and attached to 3.5 mm (diameter) × 13 mm (length) commercially pure titanium (Type IV), screw-type tapered implants with a TiUnite® acid-etched surface, and a tri-lobe internal connection (Replace Select). Abutment screws were torqued to 35 Ncm in accordance with the manufacturer's directions; the assemblies were measured and placed into a lubricated plastic ring and cap assembly (Acrylic pouring molds, Allied High Tech Products Inc. Rancho Dominguez, CA) and embedded into epoxy resin (Epoxycure, Buehler, Lake Bluff, IL) so that 3.0 mm of the implant was exposed above the resin to simulate the worst-case clinical scenario for crestal bone loss in accordance with ISO:14801 (Fig 10).

Fast fracture

Fast fracture testing was accomplished prior to cyclic fatigue testing to determine an upper limit for the amount of force necessary to cause failure. The upper limit of force was necessary to design the load-time profile for an efficient fatigue test. One specimen from each group was used for this purpose; the remaining six specimens for each group were saved for use during the cyclic fatigue testing. Fast fracture specimens were loaded monotonically at a rate of 0.5 mm/min using a servo-hydraulic test machine (Minibionix II, MTS Systems Corporation, Eden Prairie, MN) (Fig 11).

Cyclic fatigue

Specimens were subjected to a step-stress accelerated lifetime test using a servo-hydraulic test machine (Minibionix II, MTS Systems Corporation) with a 2-Hz sine wave and 0.1 min/max load ratio. The load amplitude started at 1.8 N and increased by 1.8 N every 60 cycles until fracture. Cyclic fatigue specimens were tested using the same loading fixture and environmental chamber as fast fracture specimens. Results were recorded within categories of number of cycles until failure and amount of axial force at failure. One-way ANOVA was used to identify statistical significance with regard to these two parameters. Duncan's Multiple Range Test was employed to locate the point of significance (a = 0.05). Step-stress statistical analysis (ALTA Pro 6, Reliasoft, Tucson, AZ) with a Weibull lifetime distribution, and an inverse power law load-life relation was used to estimate the median lifetime under a cyclic load with constant amplitude of 600 N for each experimental group. In addition, Kaplan-Meier analysis (SigmaPlot 10, Systat Software, Richmond, CA) was used to construct cumulative failure probability models for control, group A, and group B specimens, and a log-rank test (a = 0.05) was used to detect significant differences between these models.

Results

Number of cycles to fracture

In the control group, the mean number of cycles to fracture was 31,205 ± 2639. In experimental group A, these numbers were 38,160 ± 4292, and in experimental group B, 31,810 ± 3408. One-way ANOVA revealed statistical significance of $p = 0.0064$ among groups. Duncan's Multiple Range Test identified significance between experimental group A, and the remaining two groups. Kaplan-Meier Survival Analysis was conducted, and the log-rank statistic for the survival curves is greater than would be expected by chance; there is a statistically significant difference between survival curves ($p = 0.012$). Unadjusted p-value in control versus experimental group A was 0.0117 (critical level 0.0170); experimental group A versus experimental group B was recorded at 0.0156 (critical level 0.0253). Both of these were statistically significant. Control versus experimental group B yielded an unadjusted p-value of 0.357 (critical level 0.0500) and was not statistically significant. The survival curve for number of cycles to fracture (Fig 12) demonstrates the above results.

Amount of axial force at fracture

In the control group, a mean axial force of 932 ± 78 N was recorded at failure. Specimens in experimental group A yielded a mean axial force of 1138 ± 128 N at fracture; in experimental group B, these numbers were 949 ± 101 N. One-way ANOVA revealed a statistical significance of $p = 0.0065$ among groups. With regard to pairwise differences among groups with regard to axial force until breakage, group A required significantly greater force ($p < 0.05$) to break than either B or the control group, which were not significant from each other. The survival curve for amount of axial force at fracture reveals an almost identical finding when compared to the survival graph for number of cycles (Fig 13), with experimental group A demonstrating a statistically significant chance of survival.

A plot of the "Cycles vs. Force" was established to identify the correlation between cycles and axial force (Fig 14). A Spearman's Correlation Coefficient of 1.000 was determined.

Discussion

The presence and position of the engaging component had significant effects on the amount of axial force and number of cycles it took until specimen failure. When compared to engaging the implant next to the cantilever or not using any engaging component, engaging the implant away from the cantilever consistently required more force, and more cycles to bring the specimens to failure. The relatively small standard deviation was indicative of well-standardized specimen production.

The location of the failure is noteworthy for its consistency. The abutment screw was the component that failed in each and every specimen in all groups. In some specimens, both screws completely separated, and this resulted in total dislodgment of the prosthesis. In others, the distal screw fractured, but the specimen stayed within the confines of the assembly contained by a bent proximal screw. No visible damage

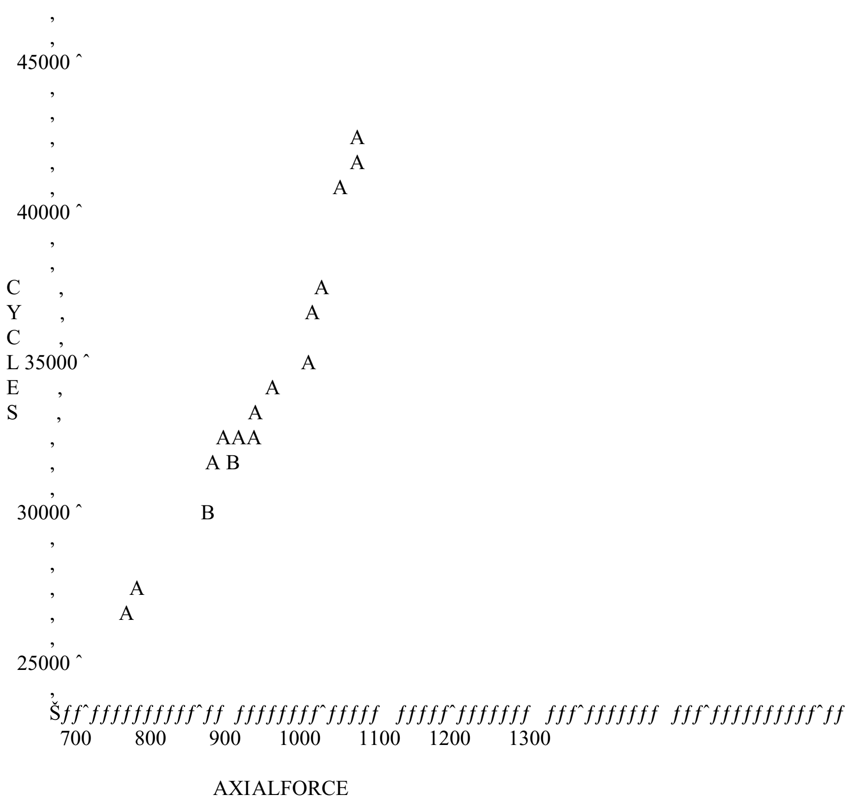

Figure 14 Cycles versus axial force correlation analysis.

was observed to any of the implant fixtures or the prosthetic specimens.

The orientation of the tri-lobe may have played a role in this failure pattern. Due to the unique design of the implants used in this investigation, the thickness of the implant wall was not uniform throughout the perimeter of the implant. As per the manufacturer's clinical protocol, one of the lobes was oriented toward the perceived buccal side of the restoration.

The axial forces measured at fracture in all three groups were substantially greater than physiologic forces seen during normal masticatory function. This can be attributed to the accelerated testing protocol, which always yields higher failure loads. It must also be pointed out that the specimens used in this investigation did not have any veneering material on the robust alloy framework. It is conceivable that the ceramic could have been the weakest component in the equation and failed long before the abutment screw. The purpose of this investigation was to test the effects of engaging abutment position on implants and related hardware. Eliminating weaker components such as porcelain and solder connections yielded more rigorous testing conditions for the implants and implant components. The forces were directed to the most undesirable point on the occlusal table of the restoration to test the effect of the cantilever. It can be argued that in the presence of a well-balanced occlusal scheme, the outcome of the investigation could have been different. Future projects can be designed to investigate the effect of different cantilever lengths when using the favorable framework design as evidenced by this investigation. Designing an adequate framework for an implant-supported prosthesis presents clinical challenges, and requires a multifaceted and complex decision-making algorithm; however, if a clinical scenario presents itself when a multiple-unit, implant-supported screw-retained prosthesis with a cantilever extension is being considered, it may be prudent to engage the implant farthest from the cantilever when designing this FDP.

Conclusions

Within the limitations of this in vitro investigation, it may be concluded that

(1) Using an engaging abutment in a screw-retained fixed cantilevered FDP provides a mechanical advantage.
(2) Engaging the implant farthest from the cantilever when designing a screw-retained cantilever FDP increased resistance to fracture of the abutment screw.
(3) When designing internally engaged implant-supported, screw-retained fixed prostheses with a cantilever component, significantly more cycles are required for failure to occur when an engaging component is used in the implant away from the cantilever.
(4) When designing internally engaged, implant-supported, screw-retained fixed prostheses with a cantilever component, significantly more force is required for early failure to occur when an engaging component is used in the implant away from the cantilever.

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