EFFECT OF IMPLANT DESIGN ON INITIAL STABILITY OF TAPERED IMPLANTS

Linus Chong, DDS, MS; Ahmed Khocht, DDS, MSD; Jon B. Suzuki, DDS, PhD; John Gaughan, PhD

Implant design is one of the parameters for achieving successful primary stability. This study aims to examine the effect of a self-tapping blades implant design on initial stability in tapered implants. Polyurethane blocks of different densities were used to simulate different bone densities. The two different implant designs included one with self-tapping blades and one without self-tapping blades. Implants were placed at 3 different depths: apical third, middle third, and fully inserted at 3 different densities of polyurethane blocks. A resonance frequency (RF) analyzer was then used to measure stability of the implants. Repeated-measures analysis of variance was used to examine the effect of implant design, insertion depth, and block density on RF. Analysis of covariance was used to examine the strength of association between RF and the aforementioned factors. In both medium-density ($P = .017$) and high-density ($P = .002$) blocks, fully inserted non-self-tapping implants showed higher initial stability than self-tapping implants. No differences were noted between the 2 implant designs that were not fully inserted. The highest strength of association was with insertion depth (standardized beta [std β] $= -0.60$, $P = .0001$), followed by block density (std $\beta = -0.15$, $P = .0002$). Implant design showed a weak association (std $\beta = -0.07$, $P = .09$). In conclusion, fully inserted implants without self-tapping blades have higher initial stability than implants with self-tapping blades. However, the association strength between implant design and initial stability is less relevant than other factors, such as insertion depth and block density. Thus, if bone quality and quantity are optimal, they may compensate for design inadequacy.

Key Words: dental implants, implant design, vibration, polyurethane

INTRODUCTION

Many studies have agreed that primary stability of a dental implant is important for the success and longevity of osseointegrated implants. Primary stability is known to be a prerequisite and a useful predictor for achieving osseointegration.[1] Adequate stability of an implant is important to allow undisturbed healing and bone formation to occur, thereby permitting optimal stress distribution from masticatory and occlusal functional loads. There are 3 main parameters for achieving primary stability: implant design, surgical technique (drill size/implant size, whether a pretapped or a self-tapped drill is used), and bone quality of the recipient site.[1] The interplay of these three parameters determines the initial stability of the implant. This study focuses mainly on implant design.

Original endosseous implants were parallel in design. However, the original design was not suitable for all applications. Other designs were introduced, including tapered implants. Tapered implants have been used to improve esthetics and facilitate implant placement between adjacent natural teeth.[2] It was initially designed especially to serve for immediate implant placement after tooth extraction. The theory behind the use of tapered implants is to provide for a

Linus Chong, DDS, MS, *is in private practice in Los Angeles, Calif.* ***Ahmed Khocht, DDS, MSD,*** *and* ***Jon B. Suzuki, DDS, PhD,*** *are with the Temple University School of Dentistry, Philadelphia, Pa. Address correspondence to Dr Khocht at the Department of Periodontology and Oral Implantology, Temple University, Maurice H. Kornberg School of Dentistry, 3223 North Broad Street, Philadelphia, PA 19140. (e-mail: akhocht@temple.edu)*
John Gaughan, PhD, *is at Temple University Biostatistics Center, Philadelphia, Pa.*

degree of compression of the cortical bone in a poor bone implant site.[3] Tapered implants distribute forces into the surrounding bone, thereby creating a more uniform compaction of bone in adjacent osteotomy walls compared with parallel-walled implants. Thus, when inserted, it creates a lateral compression of the bone.[4] The advantages can be seen especially with anatomic constraints, including ridges with concavities or narrow ridges. Cylindrical wide-bodied implants tend to run the risk of labial perforation due to buccal concavities, while the decrease in diameter toward the apical region of the tapered implant accommodates for the labial concavity.[5]

Most implant companies offer tapered implants. Nobel Biocare introduced an externally hexed, tapered titanium implant with diminishing threads in March 1997.[6] Currently, all of the major implant companies carry tapered implants: Replace Select (Nobel Biocare, Yorba Linda, Calif), Osteotite NT (Biomet 3I, Palm Beach Gardens, Fla), and Tapered Effect (Straumann, Basel, Switzerland).

Other design modifications include self-tapping implants. Self-tapping implants have been specifically designed for use in bones with poor quality (ie, Type 3 and 4 bone). Its design emphasis is placed on enhancing the primary stability of the implant. It is thought that when the self-tapping design is inserted, the denser cortical bone is compressed, thereby increasing the primary stability of the implant. The self-tapping implant also has an increased cutting characteristic. This design would help eliminate pretapping procedures in immediate implant placement resulting in improved initial implant stability.[7]

Self-tapping implants are usually designed with vertical cutting blades in the apical third of the implant. The cutting blades reduce the thread surface area and thus may minimize implant-bone contact in the apical third. It is not clear if the cutting blades influence the initial stability of the implant. This knowledge is important as it would allow for better understanding of how implant design affects implant stability and would translate clinically into a more successful outcome. The present study aims to examine the effect of self-tapping blades on initial stability in tapered implants. The hypothesis is that because self-tapping implants have fewer threads, they will have less primary stability than implants that are not self-tapping and have continuous threads throughout the implant.

Methods and Materials

For the purpose of this study, polyurethane blocks (Pacific Research Laboratories, Vashon, Wash) were used to simulate differing bone density. Two different implant designs were used: one with self-tapping blades (Biohorizons, Birmingham, Ala) and one without self-tapping blades (Nobel Biocare) as seen in Figure 1. A resonance frequency (RF) analyzer, Osstell Mentor (Ostell, Göteberg, Sweden), was used to measure implant stability.

Polyurethane blocks

Polyurethane blocks were used at different densities to simulate bone in an in vitro setting. Rigid polyurethane blocks of 13 cm $\times$ 18 cm $\times$ 4 cm were used.[8] The American Society for Testing Materials has shown that polyurethane blocks have mechanical properties simulating human bone.[9] Polyurethane is considered to be the standard material used for performing mechanical tests on orthopedic implants.[10] The blocks came from the same batch and were accurately weighed. Using the Misch classification of bone density,[11] D1 bone (high-density bone) was simulated using 50 pounds per cubic foot (pcf) polyurethane blocks, D2 bone (medium-density bone) was simulated using 40 pcf polyurethane blocks, and D3 bone (low-density bone) was simulated using 10 pcf polyurethane blocks.[12]

Implants

Twenty implants were used in the study: 10 tapered self-tapping and 10 tapered non-self-tapping. The diameter of the self-tapping implants was 4.6 mm and the diameter of the non-self-tapping implants was 4.3 mm. The length was 12 mm for self-tapping implants and 13 mm for non-self-tapping implants. The same 20 implants were used for each of the polyurethane blocks used in the study.

Resonance frequency measurements

Implant stability was measured using an Osstell Mentor.[13] The measuring technique used does not require any contact with the implant and is noninvasive. A SmartPeg is attached to the implant or abutment and is dependent on the implant manufacturer and implant diameter. The Osstell Mentor is then placed at 2 different directions, as recommended by the manufacturer.[13] An RF measurement of the peg is then taken and displayed as an implant stability quotient (ISQ) value, which is dependent on the stability of the implant. The clinical range of ISQ has been found to be normally at 50 to 80 for implant stability. Four readings were taken for each implant, and the ISQ values were then averaged.

FIGURE 1. Implant designs used in the study, self-tapping (left) and non-self-tapping (right).

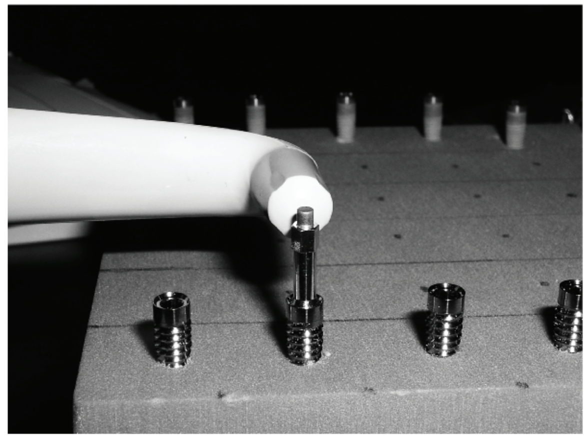

FIGURE 2. Use of Osstell machine to obtain resonance frequency reading.

Study protocol

Based on their density, the blocks were divided into 3 groups: low-density (9 blocks), medium-density (9 blocks), and high-density (9 blocks). Each density group was subdivided into 3 depth subgroups, 1/3 insertion depth (3 blocks), 2/3 insertion depth (3 blocks), and full-insertion depth (3 blocks). Thus, 27 blocks were used. The surface of each of the 27 blocks was divided into 20 equal divisions. The experiment was started with a low-density block at 1/3 insertion depth. Twenty implant insertion sites were prepared in each division on the block surface: 10 to receive self-tapping implants and 10 to receive non-self-tapping implants. Site preparation was performed as per manufacturer's recommendation for each of the 2 respective implant designs. The non-self-tapping site preparation involved 3 drills and the self-tapping site preparation involved 5 drills. The sites were prepared to receive the implant at 1/3 of their total length. Rubber stoppers were used on the cutting burs to help maintain the correct measurements. Twenty implants, 10 self-tapping and 10 non-self-tapping, were inserted in their respective sites at 1/3 of their total length. The RF was measured using the Osstell mentor device (Figure 2). Two readings in ISQ were recorded for each inserted implant. The 20 implants were then removed.

Using the same 20 implants that were used previously, the same procedures were repeated again on 2 additional blocks at 1/3 insertion depth. The same procedures were repeated again on 3 new low-density blocks at 2/3 insertion depth, and repeated again on 3 new low-density blocks at full-insertion depth. The same protocol was repeated for the medium-density blocks and the high-density blocks. Thirty duplicate RF readings were recorded for each implant design for each density block group and each insertion depth subgroup. The total of duplicate RF readings for each implant design was 270.

Statistical analysis

The data were analyzed using version 9.1 SAS software (SAS Institute, Cary, NC). The RF double readings recorded were averaged. The averaged RF data failed the test for normality so the data were transformed into normalized ranks in order to use analysis of variance (ANOVA) methods. Repeated-measures ANOVA using compound symmetry for the covariance structure was used. The dependent variable was resonance frequency. Main effects and interactions of manufacturer, depth, block density, and implant run were tested. Analysis of covariance (ANCOVA) was used to examine the strength of association between implant design and the independent variables of initial stability, block density, insertion depth, and implant run (number of times the implant was used, expressed as continuous variables).

RESULTS

Readings recorded from low-density blocks were unreliable and their data were excluded from the analysis. Only data obtained from medium-density and high-density blocks could be used.

Table 1 compares the initial implant stability of self-tapping versus non-self-tapping implant design

Table 1

Initial stability of self-tapping vs non-self-tapping implants by block density and insertion depth

Block Density	N	Insertion Depth	Self-tapping Mean (SD)	Non-self-tapping Mean (SD)	*P* Value
Medium (D2)	30	1/3	36.24 (9.20)	38.89 (11.18)	NS*
	30	2/3	55.00 (4.56)	55.91 (4.35)	NS
	30	3/3	60.60 (4.67)	63.41 (5.06)	.017
High (D1)	30	1/3	45.87 (2.80)	42.57 (4.32)	NS
	30	2/3	55.75 (4.18)	58.40 (4.18)	NS
	30	3/3	62.39 (6.27)	66.06 (5.51)	.002
All	180	All	52.64 (10.63)	54.21 (11.89)	.001

*NS indicates not significant at $P > .05$.

Table 2

Analysis of covariance evaluating the effect of implant design on initial stability combined with bone quality, insertion depth, and number of implant usage*

Dependent Variable	Independent Variables	Standard β	*P* Value
Initial stability	Block density	–0.15	.0002
	Insertion depth	–0.60	.0001
	Implant design	–0.07	.09
	Run (n times the implant was used)	0.01	NS†

*Number of observations = 180.
†NS indicates not significant at $P > .05$.

by block density and insertion depth. In medium-density blocks, fully inserted non-self-tapping implants showed higher initial stability than self-tapping implants ($P = .017$). No differences were noted between the 2 implant designs that were not fully inserted. High-density block RF readings were similar to medium-density block RF readings. Again, fully inserted non-self-tapping implants showed higher initial stability than self-tapping implants ($P = .002$). No differences were noted between the 2 implant designs that were not fully inserted. Overall comparisons between the 2 implant designs in all bone types and insertion depths showed that non-self-tapping implants have higher initial stability than self-tapping implants ($P = .001$). Thus, data from this analysis indicate that in general non-self-tapping implants have greater initial stability than self-tapping implants.

Table 2 summarizes ANCOVA analysis examining the association of implant design, block density, insertion depth, and number of runs (analyzed as continuous variables) on initial stability. Implant design, block density, and insertion depth all showed an independent relevant effect on initial stability. The highest strength of association was with insertion depth (standard beta [std β] = -0.60, $P = .0001$), followed by block density (std $\beta = -0.15$, $P = .0002$). Association with implant design was weak (std $\beta = -0.07$, $P = .09$). The number of runs (number of times the implant was used) showed no association with implant stability. Thus, implant design seems to have a less relevant effect on initial implant stability than insertion depth and block density.

Discussion

This study attempted to investigate the effect of self-tapping design on initial stability of tapered implants in polyurethane bone blocks. The findings indicate that tapered implants without self-tapping blades have higher initial stability than implants with self-tapping blades.

It is of interest to note that no difference in initial stability was found between self-tapping and non-self-tapping implant designs in both medium-density and high-density blocks at the 1/3 and 2/3 insertion depths. This indicates that the overall higher initial stability of the non-self-tapping implant at full insertion depth is not solely associated with the design of the apical third of the implant but is most likely associated with the design of the entire implant surface. The non-self-tapping implant has a higher number of threads than the self-tapping implant, thus increasing the surface area in contact with adjacent block/bone walls. In the self-tapping implant design, the self-tapping blades present in the apical third of the implant replace implant threads and minimize the contact surface area of the implant. Interestingly, the lateral compression of material/bone associated with self-tapping thread design did not enhance the primary stability of the self-tapping design used in this study.

Other factors that were not investigated in this study include differences in the osteotomy preparation protocols and surface texture of the 2 different implants. The different implant designs require different osteotomy preparation protocols. The non-self-tapping osteotomy preparation protocol used a fewer sequential cutting burs. Also, the 2 implants used had different surface textures: the non-self-tapping implant had a roughened surface and the self-tapping had a smooth surface. Thus, the higher initial stability of the non-self-tapping implant may be due to a combination of factors, including more threads on the implant surface, a roughened surface, absence of self-tapping blades, and a conservative osteotomy drilling sequence.

The American Society for Testing Materials has validated polyurethane blocks as having mechanical properties simulating human cancellous bone.[9] Polyurethane has been used as a standard material for performing mechanical tests for orthopedic implants. This study uses polyurethane blocks at different densities to simulate the different bone densities in humans according to the Misch classification of D1, D2, and D3 bone.[12]

In this study, D3 polyurethane blocks could not be included because the blocks did not have the same dampening effect as the maxilla and mandible. Stability at 1/3 depth could not be measured because of significant vibration of the D3 block once resonance frequency was emitted. Placing the blocks on sticky pads and holding them in place did not decrease the amount of vibration; therefore, results could not be determined. However, the different densities were consistent with what was expected in that an increase in stability of the implants was found as the density of the bone blocks increased.

In the past, different tests have been done to test stability of implants, including percussion, finger pressure, radiographs, Periotest (Medizintechnik Guldne, Modautal, Germany) RF analysis, and insertion torque. According to Andreaza da Cunha et al,[14] RF analysis and insertion torque methods appear to be most efficient and have fewer contraindications.

One way of quantitatively evaluating implant stability and osseointegration is through RF analysis. Meredith et al[15] introduced a noninvasive method for quantitatively measuring implant stability through RF measurements related to the effective length of an implant above the level of the bone. This method requires placement of an electronic transducer onto the implant head or prosthetic abutment with a retaining screw. A low-voltage current, undetectable by the patient, is then passed through the transducer. Original measurements were made in Hertz. These measurements are then calibrated for each transducer and converted into ISQ units by a computer. A decrease in the RF value is related to a decrease in stiffness, which can indicate a potential for failure. Thus, RF analysis can be used to show differences between successful implants and clinical failures in order to predict conditions that predispose an implant to fail.[16]

The RF analysis value is also a measurement of stiffness of the implant/transducer/bone complex and is influenced by the length of the implant above the bone crest.[16] It has been used for to determine changes in implant stability in bone healing or failure of implants to integrate. One way of measuring RF is through the use of an Ostell RF analyzer. The values are calculated from peak amplitude, called the ISQ value, which is measured on a scale from 1 to 100. A high ISQ value indicates high stability, and a low ISQ value shows low stability.[13]

Several studies have used the Osstell Mentor in obtaining RF analysis measurements of implant stability. In addition, RF analysis has been noted as an objective method in determining implant stability.[17] It can also be applied as a predictor of implant success. In this study, a standardized measuring protocol was developed to ensure obtaining consistent results. Balshi et al[18] recommended that a parallel orientation of the transducer should be used for all clinical RF analysis measurements. The ISQ level is often influenced by the orientation of the transducer to the axis of the alveolar ridge, where it has been found that perpendicular measurements have ISQs equal to or higher than their parallel measurements.[18] In the current study, 2 readings of both orientations were taken and then averaged.

A comparison of different implant designs has been examined in the past. O'Sullivan et al[19] compared the primary stability characteristics of five different implant designs, specifically the standard Branemark, Mark II self-tapping implant and Mark IV self-tapping tapered implant (Nobel Biocare AB, Goteburg, Sweden), Osseotite (Biomet 3I), and Tioblast implant (AstraTech AB, Molndahl, Sweden). The study demonstrated higher RF analysis and insertion torque values for tapered implants than for nontapered implants, suggesting increased stability in tapered implants.[19] In looking at the Branemark system and the Mark II, III, and IV implants, Glauser et al[20] also found higher RF analysis values for tapered implants (Mark IV) than for nontapered implants (Branemark, Mark II and III). The current in vitro study extends the findings of these 2 reports by examining the effect of the self-tapping design on the initial stability of the tapered implant.

In conclusion, the current study demonstrates that the design of the entire implant surface, not just the apical third, is important for initial implant stability. The non-self-tapping design tested in this study had greater initial stability when fully inserted. However, the findings also indicate that insertion depth and bone density have stronger association with initial stability than implant design. Thus, when an adequate amount of high-quality bone surrounds the implant, it may compensate for design inadequacy.

Acknowledgments

This study was sponsored in part by a 2007 American Academy of Implant Dentistry Grant and a 2007/2008

Temple University School of Dentistry Research Incentive Grant.

References

1. Meredith N. Assessment of implant stability as a prognostic determinant. *Int J Prosthodont*. 1998;11:491–501.

2. Shapoff C. Clinical advantages of tapered root form dental implants. *Compend Contin Educ Dent*. 2002;23:42–58.

3. O'Sullivan D, Sennerby L, Meredith N. Influence of implant taper on the primary and secondary stability of osseointegrated titanium implants. *Clin Oral Implant Res*. 2004;15:474–480.

4. Glauser R, Sennerby L, Meredith N, et al. Resonance frequency analysis of implants subjected to immediate or early functional occlusal loading. Successful vs. failing implants. *Clin Oral Implant Res*. 2004;15:428–434.

5. Garber DA, Salama H, Salama M. Two stage versus one stage—is there really a controversy? *J Periodontol.* 2001;72:417–421.

6. Callan DP, Hahn J, Hebel K, et al. Retrospective multicenter study of an anodized, tapered, diminishing thread implant: success rate at exposure. *Implant Dent*. 2000;9:329–333.

7. Olsson M, Friberg B, Nilson H, Kultje C. MkII—a modified self-tapping Branemark implant: 3-year results of a controlled prospective pilot study. *Int J Oral Maxillofac Implants*. 1995;10:15–21.

8. Sawbones Worldwide. Products and services. Pacific Research Laboratories Ltd., Vashon, Wash. http://www.sawbones.com. Accessed March 11, 2008.

9. American Society for Testing Materials. Standard specification for rigid polyurethane foam for use as a standard material for testing orthopaedic devices and instruments. Report: F1839-97. http://www.astm.org. Accessed June 11, 2008.

10. Sunit P, Mahon A, Green S, McMurtry I, Port A. A biomechanical study comparing a raft of 3.5 mm cortical screws with 6.5 mm cancellous screws in depressed tibial plateau fractures. *Knee.* 2006;13:231–235

11. Misch CE. *Contemporary Implant Dentistry*. 2nd ed. St Louis, Mo: Mosby Co; 1999:372–379.

12. Orchid-Orthopedics. Markets/dental. http://www.Orchid-Orthopedics.com. Accessed June 11, 2008.

13. Osstell Mentor. Osstell. Göteberg, Sweden. http://www.asstell.com. Accessed March 12, 2008.

14. Andreaza da Cunha H, Francischone CE, Filho H, Gomes de Oliveira R. A comparison between cutting torque and resonance frequency in the assessment of primary stability and final torque capacity of standard and TiUnite single-tooth implants under immediate loading. *Int J Oral Maxillofacial Implants*. 2004;19:578–585.

15. Meredith N, Alleyne D, Cawley P. Quantitative determination of the stability of the implant tissue interface using resonance frequency analysis. *Clin Oral Impl Res*. 1996:7:261–267.

16. Meredith N, Book K, Friberg B, Jemt T, Sennerby L. Resonance frequency measurements of implants stability in vivo. A cross-sectional and longitudinal study of resonance frequency measurements on implants in the edentulous and partially dentate maxilla. *Clin Oral Impl Res*. 1997:8:226–233.

17. Becker W, Sennerby L, Bedrossian E, Becker B, Lucchini JP. Implant stability measurements for implants placed at the time of extraction: a cohort, prospective clinical trial. *J Periodontol*. 2005;76:391–397.

18. Balshi SF, Allen FD, Wolfinger GJ, Balshi TJ. A resonance frequency analysis assessment of maxillary and mandibular immediately loaded implants. *Int J Oral Maxillofac Implants*. 2005;20:584–594.

19. O'Sullivan D, Sennerby L, Meredith N. Measurements comparing the initial stability of five designs of dental implants: a human cadaver study. *Clin Implant Dent Rel Res*. 2000;2:85–92.

20. Glauser R, Portmann M, Ruhstaller P, Gottlow J, Scharer P. Initial implant stability using different implant designs and surgical techniques. A comparative clinical study using insertion torque and resonance frequency analysis. *Appl Osseointegration Res*. 2001;2:6–8.