A Comparison of Two Methods of Enhancing Implant Primary Stability

Dominic O'Sullivan, BDS, FDS RCS, PhD;* Lars Sennerby, DDS, PhD;† Daryll Jagger, BDS, FDS RCS, PhD;‡ Neil Meredith, BDS, PhD, FDS RCS§

ABSTRACT

Background: Surgical technique and implant design have an effect on the primary stability of oral implants, which in turn increases resistance to implant micromotion during healing.

Purpose: This study was designed to compare the parameters associated with implant insertion using two different methods of enhancing implant primary stability and to identify any relationship between these parameters and changes in the stability of implants during the initial 6-month healing period following implant insertion. A comparison was made between two methods of enhancing primary implant stability: method 1, standard Brånemark System® implants (Nobel Biocare AB, Gothenburg, Sweden) inserted with a technique designed to enhance primary stability, and method 2: Brånemark Mk IV implants (Nobel Biocare AB) inserted according to the manufacturer's instructions.

Materials and Methods: Thirteen patients were selected for inclusion in the study. A total of 42 implants were placed. Insertion torque data were recorded, and bone quality at the implant site was assessed at implant insertion. Resonance frequency analysis measurements were taken at implant insertion as well as at second-stage surgery 6 months later.

Results: A statistically significant difference was recorded between the mean maximum insertion torque for type 4 bone and bone types 2 and 3. No significant difference was recorded between bone types 2 and 3. A significantly lower resonance frequency value was seen for standard implants placed into type 4 bone ($p < .05$). Across all implant types a significant difference in the energy required when inserting implants into type 4 bone and bone types 2 and 3 was seen. A significantly lower mean energy requirement was seen between the Mk IV implants placed into type 4 bone and the other combinations of implant types and bone.

Conclusions: Within the limitations of this study, the results agree with the manufacturer's claim that when compared with standard implants, the design of the Mk IV implant increases implant primary stability with a reduction in the energy imparted into the bone at the implant site.

KEY WORDS: implant design, primary stability, resonance frequency analysis

The successful preparation of bone for the reception of an oral implant should allow close apposition between the bone and the implant surface. When an implant is placed into bone, the aim is to drill a hole in the bone into which the implant will be placed so that the implant surface and bone surfaces are closely apposed. However, in practice this cannot always be achieved with current surgical drilling and tapping procedures. Owing to changes in surgeon's hand position and the chatter and movement of the drill in the handpiece, the hole will often deviate from the ideal. A study using sheep tibiae[1] suggested that up to 30% of implant preparations showed such deviations. This mismatch between the implant site and the implant creates gaps between the implant surface and the bone; if a load is applied to the implant before bone has been able to grow into this space, micromotion may occur. In a severe case motion may be detectable clinically.

*Clinical lecturer, Bristol Dental School, Bristol, United Kingdom; †professor, Department of Biomaterials, Institute for Surgical Sciences, University of Gothenburg, Gothenburg, Sweden; ‡consultant senior lecturer, Bristol Dental School, Bristol, United Kingdom; §professor, Leeds Dental Institute, University of Leeds, Leeds, United Kingdom

Reprint requests: Dominic O'Sullivan, BDS, FDS RCS, PhD, Bristol Dental School, Lower Maudlin Street, Bristol BS1 2LY, UK; e-mail: d.j.osullivan@bris.ac.uk

However, it has been suggested that implants may more commonly show signs of subclinical mobility, which may have an effect on implant integration.[2,3] Previous studies all showed that micromotion between the implant and the bone promotes the formation of a soft tissue capsule around the implant.[4–8] A degree of motion appears to be tolerated before soft tissue formation occurs. The absolute value of tolerated differential movement of the implant and bone surfaces has yet to be fully defined for intraoral implants, and certainly such a precise level of motion is impossible to control in vivo. In practice surgeons aim for the maximum stability achievable. The stability of an implant immediately after surgical placement is defined as primary implant stability. Primary implant stability is affected by several factors, including bone quality, bone quantity, implant geometry, and the relationship between the pilot hole/tapped channel and the implant's diameter.[9] Efforts may be made to modify the bone quality and quantity by the use of bone grafts or augmentation materials, but for the majority of implant insertions, the quality of the bone at the implant site is a parameter over which the surgeon has little control. The design of the implant and the surgical technique used are the two factors over which the surgeon may be able to exercise choice and influence to improve the implant's primary stability.

Recent work, particularly in the field of resonance frequency, has helped to expand the understanding of primary implant stability and its relationship with secondary implant stability and clinical outcome.[9–16] The use of resonance frequency analysis (RFA) to quantitatively examine the implant-tissue interface has been described.[12] Several studies have reported that the RFA value of the implant/transducer complex is related to the height of the implant above the bone crest and the stability of the implant-tissue interface as determined by the absence of clinical mobility.[13–15]

Repeated RFA measurements made on healing implants in rabbit tibiae demonstrated an initial rise in value and a leveling off after 43 days.[15] Histomorphometric analysis of bone formation around the implant during healing has demonstrated a correlation between bone formation adjacent to the implant and a rise in RFA value relating to increased implant stability. It is possible that the healing process around the implant after insertion alters the implant's initial primary stability. It has also been suggested that primary stability is determined by (1) the density and quantity of the bone at the implant site, (2) the surgical technique used, and (3) the design of the implant.[17] This study suggested that secondary stability, which is seen after the healing period, is primary stability with a further gain in stability because of bone formation around the implant. By extension, bone loss during healing or the formation of a fibrous tissue capsule may cause a reduction in the initial primary stability value, indicating failure of the implant. It may be expected that an implant with an initially high stability may exhibit a loss of stability because of the net activity of resorptive cells during a remodeling phase whereas stability will be enhanced by the formation of new bone in close contact with the implant surface. The interaction between these factors is complex and as yet unquantified. A correlation between the removal torque of titanium implants and the amount of compact bone surrounding them when they are placed into the tibia and cancellous bone adjacent to the knee joints of rabbits has been reported.[18] The cortical bone appeared to provide improved implant support in the immediate postoperative period. With time the stability of the implants placed into the cancellous bone reached the same levels, but the implants were relatively vulnerable in the immediate postoperative period. It has also been suggested that the pre-tapping of a threaded bone channel at placement be omitted in cases in which minimal cortical bone is present, in order to induce compression in the interfacial bone and enhance stability.[19]

Over the years research into oral implants has led to the development of the standard Brånemark screw-form commercially pure (CP) titanium implant, which has become one of the most widely studied implant designs.[20,21] Brånemark Standard implants (Nobel Biocare AB, Gothenburg, Sweden) were not designed to be self-tapping and have no effective cutting facets. To enhance the stability of these implants at initial surgery, many surgeons placed these implants without using a surgical tap to prepare a threaded channel in the bone. This technique allows the placement of the implant in slight compression within the bone. In theory this compression enhances implant primary stability by developing circumferential or hoop stresses within the bone at the zone of the bone-implant interface. This method of enhancing primary implant stability is

based purely on intuitive reasoning, and the exact effect of this technique has not been quantified.

The Brånemark System® Mk IV implant (Nobel Biocare AB) was designed to create differential stresses within the bone at the implant site. The design attempts to induce circumferential stresses of greater magnitude in the cortical bone as compared to the trabecular bone, to enhance the implant's primary stability in a manner similar to the technique described above for standard implants. The implant design was developed with the aim of inducing the greatest stresses within the denser cortical bone layer; this was achieved by giving the Mk IV implant a slightly tapered profile. The Mk IV was also given a double start thread, which reduced insertion time. The manufacturers claimed that the taper induced compression within the cortical bone while the double thread reduced the thermal energy generated at the bone-implant interface, owing to the reduced insertion time. The reduced thermal energy transmitted to the bone should minimize osteogenic bone cell damage due to heat generation. No evidence is available in the literature to support this claim.

The aim of this study was to compare selected parameters associated with implant insertion using two different methods of enhancing implant primary stability and to identify any relationship between these parameters and changes in the stability of each implant during the initial 6-month healing period following implant insertion.

MATERIALS AND METHODS

Ethical Approval

Ethical approval for this study was gained from the Local Research Ethics Committee.

Patient Selection and Implant Placement

Thirteen patients were selected for inclusion in the study. The patients were chosen as a representative sample of patients who were referred to a university dental hospital and school to be provided with dental implants. The mean age of the patients was 39 years, and the ratio of male to female patients was 5:8. Two types of implant were included in this study: Brånemark Standard implants and Mk IV implants. Details of the type and length of implant and bone quality at the implant site are presented in Table 1. A total of 42 implants were placed, 38 in the maxilla and 4 in the mandible. To reduce interoperator variability one operator placed all of the implants. Standard surgical techniques were used to prepare the surgical sites. Full-thickness mucoperiosteal flaps were raised while the patients were under local anesthesia (Xylocaine® 2% with adrenaline, 1:80,000, Dentsply Pharmaceutical, Weybridge, Surrey, UK). A 2 mm roundheaded guide drill was used first, to locate the implant position on the cortical bone surface. A 2 mm diameter twist drill was used (under profuse isotonic saline irrigation) to prepare the initial full-depth channel at the implant site. A pilot drill was then used to enlarge the diameter of the most coronal portion of the channel, and the channel was subsequently enlarged to 3.15 mm in diameter with a twist drill. Profuse irrigation with sterile isotonic saline was used at each drilling stage. Standard and Mk IV implants were inserted at the same slow rotational speed under profuse sterile saline irrigation. All drilling and implant insertion procedures were carried out with the Osseocare™ drill controller (Nobel Biocare AB).

Data Collection

At implant placement, data were recorded onto a "smart card" by the Osseocare drill controller. Insertion torque was derived from the current taken by the motor during implant insertion and recorded in newton-centimeters at every 90° rotation of the implant. Processing was performed internally in the Osseocare unit, and the insertion torque and degree of rotation were recorded as a compressed American Standard Code for Information Exchange (ASCII) text file for each implant.

Calibration of the Osseocare Unit

Preliminary studies were undertaken to calibrate the Osseocare unit as few data were available owing to its only having recently been introduced at the time of this study. Although the manufacturer claimed that the unit was factory calibrated to calculate torque and rotation, it was felt necessary to test the unit against a laboratory standard. To test the insertion torque accuracy of the unit, the handpiece was connected to a surgical tap placed into the grips of a UK National Measurement Accreditation Service (NAMAS)-certified new Tohnichi torque gauge (Tohnichi MFG, Tokyo, Japan). The Osseocare unit allows the operator to select preset maximum insertion torque values. When the foot pedal is activated, the handpiece shaft rotates at the selected

TABLE 1 Implant Distribution by Length of Implant and Bone Quality at Implant Site

	Standard Implant				Mk IV Implant			
Bone Quality	Total	10 mm	13 mm	15 mm	Total	10 mm	13 mm	15 mm
2	15	2	11	2	—	—	—	—
3	9	2	5	2	9	0	5	4
4	4	0	2	2	5	0	5	0

speed (high or low) until the maximum insertion torque is reached, at which point the unit shuts off. For calibration each maximum insertion torque value was selected, and the foot pedal was activated until the preset value was reached and the shutoff mechanism was triggered. The data collected on the Osseocare smart card were then compared with the value recorded on the Tohnichi torque gauge. This procedure was repeated five times at each preset value. The results of the Osseocare torque calibration are summarized in Figure 1. Although the manufacturer of the Osseocare unit claimed that data relating to the rotation of the working end of the handpiece were derived from recording the commutator pulses from the motor, the exact method used was considered commercially sensitive at the time of this study, and it was therefore felt necessary to calibrate this unit against a laboratory standard encoder. To calibrate the handpiece for rotation, the handpiece was connected to an implant firmly seated into a factory-certified commercially available optical rotary encoder (HEDS-550S, Hewlett-Packard Ltd, Bracknell, Berkshire, UK). The encoder was connected to the input channel of a data acquisition card (MIO A10-16XE 50, National Instruments Ltd, Newbury, Berkshire, UK), and the encoder signal was then conditioned and recorded with a custom-programmed virtual instrument and commercially available software (Labview® 5.1, National Instruments Ltd, Newbury, Berkshire, UK). Encoder data were converted into degrees of rotation and were saved as an ASCII text file to the hard drive of a personal computer (PC). The foot pedal was activated to rotate the implant in the encoder by approximately one full revolution, and the data were logged simultaneously with the Osseocare and the encoder/PC. The results of the rotation calibration are shown in Figure 2.

Bone Quality Assessment

At each implant site the operator made an assessment of the bone quality according to the scoring system

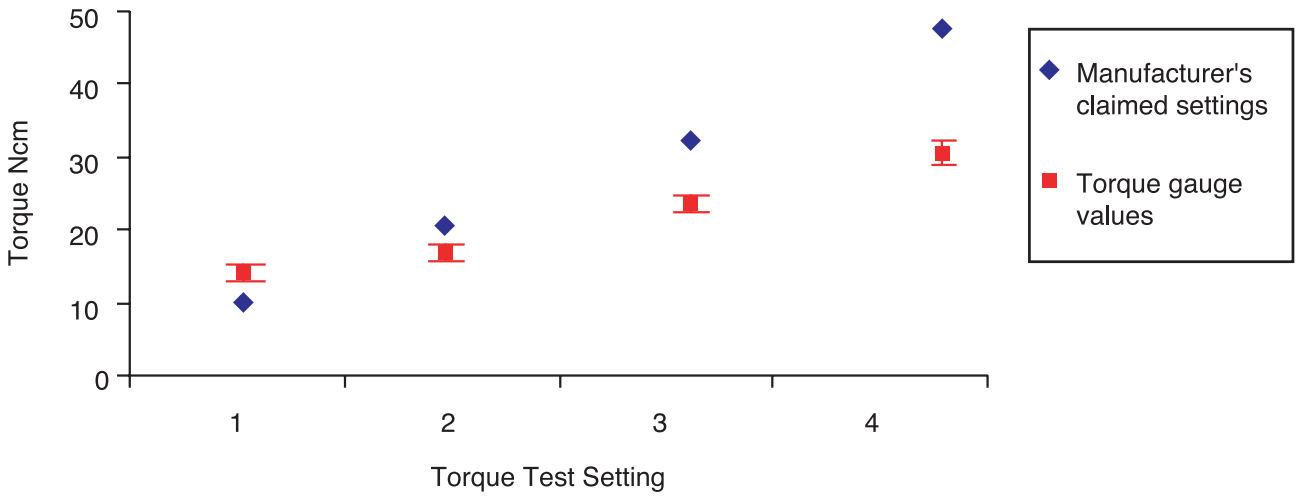

Figure 1 Osseocare torque calibration results. Mean values with 95% confidence intervals are shown where appropriate ($n = 20$).

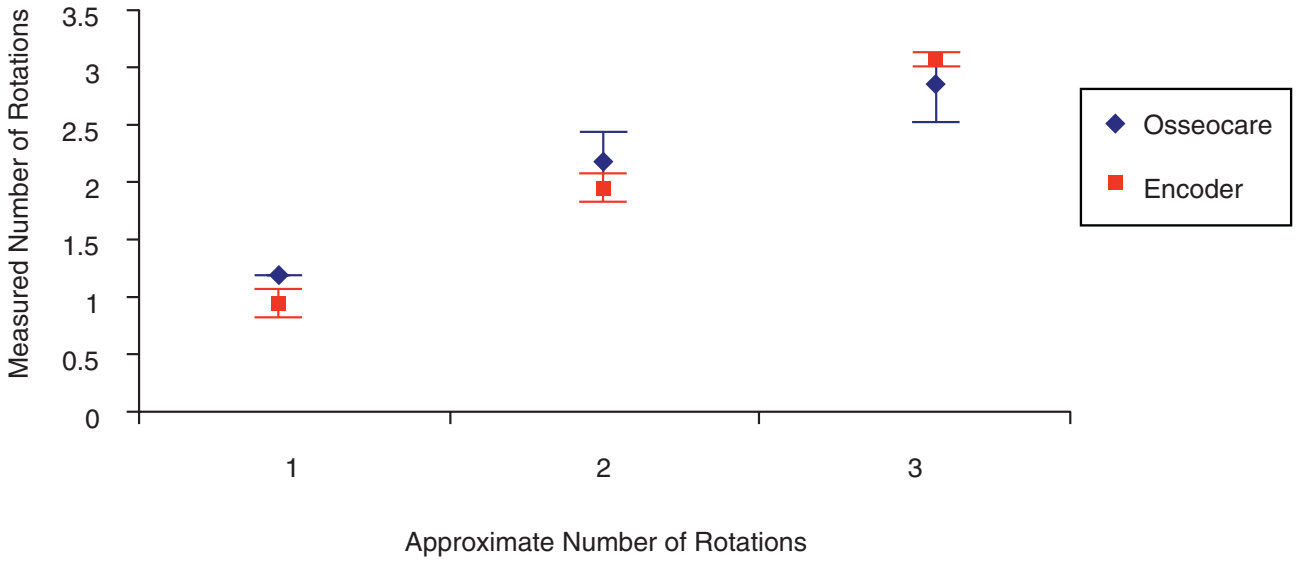

Figure 2 Osseocare rotation calibration results. Mean values and 95% confidence intervals are shown (n = 15).

devised by Lekholm and Zarb.[22] This assessment was based on the appearance of the site on preoperative radiographs and on the resistance felt to cutting and tapping when preparing the implant site and placing the implant. The operator was blinded to any of the insertion torque data, and the bone quality assessment was made prior to the resonance frequency test.

Resonance Frequency Analysis

RFA readings for each implant were taken at implant placement and 6 months post insertion, after healing had taken place. After implant placement resonance frequency measurements were made according to a previously described method.[9]

Evaluation of Insertion Torque Peaks

The maximum insertion torque peak was taken as the maximum torque value recorded during implant insertion. The slope of the final peak relating to the maximum insertion torque value was examined by taking a tangent from the insertion torque plot at this point and calculating the slope of the tangent.

Calculation of Energy Required during Implant Insertion

The energy required to insert an implant was determined by plotting the insertion torque in Newton-metres (Nm) against the angular displacement of the implant in radians. The area under the resulting curve represents the energy used (in joules).[23]

Statistical Comparison

Statistical comparison was done with analysis of variance. When a significant difference was indicated, the Bonferroni multiple comparison test was carried out with the significance set at p = .05.

RESULTS

Insertion Torque

The mean maximum insertion torque values generated during implant insertion for each of the three bone qualities (2, 3, and 4) are shown in Table 2. Mean

TABLE 2 Insertion Torque Data for Each Bone Quality

	Bone Quality		
Torque Measurement	2	3	4
Mean maximum insertion torque at implant insertion (Ncm)	26.6	25.9	13.3
95% confidence interval	2.3	2.6	5.3
Slope of tangent to maximum insertion torque peak (y/x)	18.2	14.0	9.1
95% confidence interval	1.8	2.0	3.4

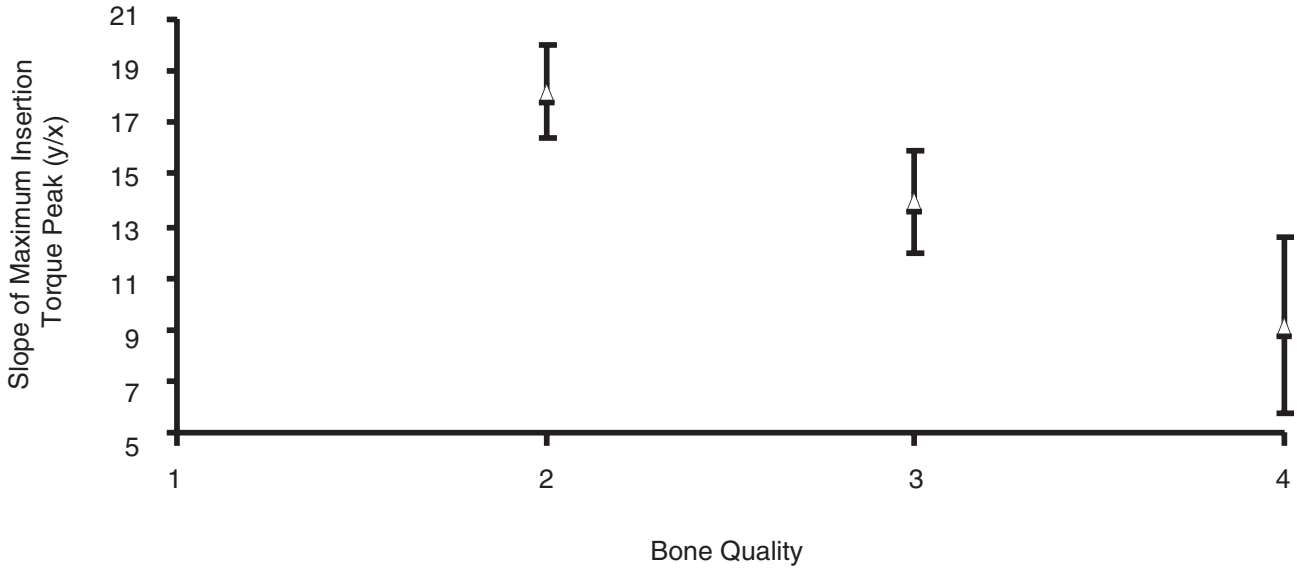

Figure 3 Slope of the tangent to the maximum insertion torque peak at implant insertion. Mean values are shown with 95% confidence intervals (n = 42).

values with 95% confidence intervals are shown for all implants placed. A significant difference (p = .05) was identified between the mean maximum insertion torque values for type 4 bone and for bone of types 2 and 3. No significant difference was recorded between type 2 bone and type 3 bone. Figure 3 shows the slope of the tangent of the maximum insertion torque peak, with 95% confidence intervals, for each bone type and all implants placed. The difference between type 4 bone and type 2 and type 3 bone was statistically significant at p = .05, and no statistically significant difference was recorded between type 2 and type 3 bone.

Table 3 lists the mean maximum insertion torque values, with 95% confidence intervals, for each implant type. Values for bone of types 3 and 4 only are shown because Mk IV implants are contraindicated for type 2 bone, and therefore no comparison could be made. No statistically significant difference was seen between the implant types for each bone quality. Although there was an apparent difference between the mean values for standard and Mk IV implants in type 4 bone when compared to type 3 bone, the difference was not statistically significant (p = .06). Table 3 also shows the slope of the tangent of the maximum insertion torque peak (calculated as y/x) for bone types 3 and 4 and for each implant type. No significant difference was seen between each of the two implant types in either type 3 or type 4 bone.

Energy required during implant insertions is indicated in Table 4. A statistically significant difference was seen between those implants placed into type 4 bone and those placed into bone of types 2 and 3. A significant difference (p = .05) was seen between Mk IV

TABLE 3 Insertion Torque Data for Each Implant Type

	Implant Type	
	Standard	Mk IV
Bone Quality 3		
Mean maximum insertion torque at implant insertion (Ncm)	24.7	29.3
95% confidence interval	3.3	2.8
Mean slope of tangent to maximum insertion torque peak (y/x)	14.5	14.1
95% confidence interval	2.3	6.3
Bone Quality 4		
Mean maximum insertion torque at implant insertion (Ncm)	14.0	12.3
95% confidence interval	9.0	2.8
Mean slope of tangent to maximum insertion torque peak (y/x)	9.1	9.2
95% confidence interval	6.4	0.7

TABLE 4 Energy Required during Implant Insertion for Each Implant Type

	Implant Type	
	Standard	Mk IV
Bone Quality 2		
Mean energy required (J)	11.15	—
95% confidence interval	1.74	—
Bone Quality 3		
Mean energy required (J)	11.18	7.39
95% confidence interval	2.74	1.19
Bone Quality 4		
Mean energy required (J)	6.43	3.23
95% confidence interval	2.48	1.28

implants placed into type 4 bone and other combinations of implant and bone types. No other statistically significant differences were noted.

Resonance Frequency Analysis

The mean RFA values at implant placement for bone quality types 3 and 4 and for each implant type, together with 95% confidence intervals, are shown in Table 5. A statistically significant difference was seen between standard implants placed into type 4 bone and the other combinations of implant and bone types. Mean RFA values at second-stage surgery after 6 months of healing for each implant type placed into type 3 and type 4 bone are presented in Table 6. As for results related to the review following the 6-month healing period, no statistically significant difference was seen.

TABLE 5 Mean Resonance Frequency Analysis Values at Implant Insertion for Each Implant Type

	Implant Type	
	Standard	Mk IV
Bone Quality 2		
Mean resonance frequency (kHz)	6.04	—
95% confidence interval	0.14	—
Bone Quality 3		
Mean resonance frequency (kHz)	6.16	6.18
95% confidence interval	0.13	0.30
Bone Quality 4		
Mean resonance frequency (kHz)	5.30	5.96
95% confidence interval	0.20	0.01

TABLE 6 Mean Resonance Frequency Analysis Values for Each Implant Type at 6-Month Review

	Implant Type	
	Standard	Mk IV
Bone Quality 2		
Mean resonance frequency (kHz)	5.90	—
95% confidence interval	0.2	—
Bone Quality 3		
Mean resonance frequency (kHz)	5.96	5.96
95% confidence interval	0.1	0.2
Bone Quality 4		
Mean resonance frequency (kHz)	5.60	5.86
95% confidence interval	0.8	0.2

DISCUSSION

Previous studies have reported that primary implant stability is of great importance when placing oral implants in bone and that efforts should be made to maximize primary stability. The surgical techniques used in this study were designed to maximize primary implant stability. It would have been interesting to compare the methods used to enhance primary stability with the traditional use of a surgical tap prior to implant insertion. In our opinion, however, this might have affected the clinical outcome for some of the implants. Quantitative data relating to the enhanced stability gained from not using a surgical tap prior to placing a nontapping implant are lacking even though that procedure has become an accepted part of current implantation surgery.

Calculating the energy used during implant insertion is complex. A technique using insertion torque data to calculate the energy used in cutting a prescribed unit of bone has been described.[24] In this study a number of assumptions were made regarding the energy used in the development of friction between the implant and the bone and in the "shiver packing" of bone chips into the cutting flutes and surface irregularities of the implant's surface. The effects of differing implant designs and variations in the pilot

hole/implant ratio on the measured insertion torque were not, however, taken into account. In this study the overall energy used during the placement of the implant as a guide to the relative differences in energy imparted to the bone at the implant site during implant insertion was considered. However, the overall energy used during insertion is an overestimate of the energy imparted to the bone since energy is also lost in the generation of heat within the handpiece, in the generation of noise, and to friction between the components of the handpiece and motor. For the purposes of this study, it was assumed that these parameters were relatively consistent and were unaffected by the type of implant placed although it was accepted that they might increase the variability between the readings.

If implant type is not taken into consideration (and taking all of the implants placed into account), maximum insertion torque appears to be a useful indicator of bone quality type 4. Although there is no statistical difference between type 2 and type 3 bone, a difference was noted in this study between types 2 and 4 and between type 3 and type 4 bone quality. The relationship between bone quality and insertion torque was investigated by Friberg and colleagues.[25] A significant correlation between cutting resistance (calculated by using the Johansson and Strid model) and bone density as well as between cutting resistance and the bone area of postmortem jaws was reported. A correlation between the maximum cutting torque values and assessed bone density scores in vivo was also noted.

Peak insertion torque can relate to a number of situations clinically. The maximum torque value generated may be due to the implant flange's impinging on the crestal cortical bone, the implant ''bottoming out'' at the base of the prepared bone channel, the engagement of a lower cortical bone layer by the apical portion of the implant, the generation of friction as the full length of the implant inserts into bone, or the resistance of interfacial bone to local compression in a tapered implant. The interrelation of these factors may explain why there does not appear to be a clear relationship between the maximum insertion torque values and the Lekholm and Zarb scores in this study. However, the slope of the tangent to the maximum insertion torque peak does appear to show a correlation to bone quality. Implants placed into type 2 bone generated a steeper slope than did those placed into type 4 bone. This difference is statistically significant ($p = .05$). This has not been previously reported in the literature and appears to be relatively unaffected by the design of the implant. From a clinical viewpoint, the Lekholm and Zarb classification has always been seen as highly subjective, and there has been little investigation into the extent to which inter- and intraoperator variability affects bone quality assessment. Both implant types in this study compress the interfacial bone significantly during insertion. The resistance of the bone to this compression generates a rapid rise in insertion torque. This contrasts with the insertion torque profile seen when a standard Brånemark implant is placed after use of the surgical tap. The slope of the graph represents the rate of application of energy to the system.

As with the maximum insertion torque value, no significant difference in mean RFA values was found between bone qualities 2 and 3 although a significant difference was observed between type 3 and type 4 bone. The similarity between RFA values for implants placed into type 2 and those placed into type 3 bone indicates that the primary stability of these implants is close to the maximum achievable under the clinical conditions of this study. The significant drop in RFA value between bone types 2 and 3 and bone type 4 may indicate that the techniques used to maximize primary implant stability in type 4 bone are unable to achieve the stability achievable in type 2 and type 3 bone. This mirrors the lower clinical implant success rate reported by a number of authors when implants are placed into type 4 bone when compared with implants placed into type 1, 2, or 3 bone.[26–28] The differences in RFA values were no longer significant 6 months after implant placement. This reflects a slight decrease in the mean RFA value in type 2 and type 3 bone and a slight increase in the mean RFA value for type 4 bone. This is in agreement with similar trends seen in recent studies relating to RFA.[29,30] Implants placed into dense bone appear to undergo minimal bone remodeling at the neck of the implant, which lowers the RFA value by increasing the effective length of the implant/transducer complex above bone. Bone apposition onto the surface of the implant during the healing period increases the stability of the implants placed into bone of poorer quality, thereby raising the mean RFA value.[31] Similar apposition occurs with implants that are placed into denser bone, but this apposition does not appear to significantly alter stability during the

healing period as the implants had an initially high primary stability.

No significant differences in insertion torque slope were found between implant types and bone qualities. This appears to support the suggestion that the rates of application of energy to the bone in both systems are similar and that neither implant system generates an insertion torque profile significantly different from that of the other.

At implant placement, the mean RFA values for each implant type in type 3 quality bone were similar, and no significant difference was seen between standard and Mk IV implants. There was no significant difference between Mk IV implants placed in type 3 and in type 4 bone. This is possibly an indication that the primary mechanical stability of Mk IV implants is less affected by bone quality. However, a significantly lower mean RFA value was obtained from the standard implants placed into type 4 bone than from those placed into type 3 bone. This may suggest that placing standard implants without prior tapping is not as effective at maintaining a high primary stability. The stability of the standard implant is perhaps more affected by the quality of bone at the implant site.

At second-stage surgery, as at implant insertion, no significant difference was seen between the mean RFA values for standard implants and Mk IV implants in type 3 bone. In type 4 bone no significant difference was seen between standard implants and Mk IV implants. This is perhaps due to an increase in the mean RFA value for the standard implants during the healing period. This finding is in accordance with those of other studies evaluating changes in RFA values between implant insertion and second-stage surgery.[14,29,30] The increase in the mean RFA value for the standard implants during the healing period has been attributed to the deposition of bone at the bone-implant interface, which can increase local support for the implant and thereby increase its stability.

With regard to the energy expended during implant insertion, no significant difference between the energy used to insert a standard implant into type 2 bone and that used to insert a standard implant into type 3 bone was recorded. However, a difference was seen between the energy required to insert a standard implant into type 4 bone and that required for insertion into the other two bone types. A difference was also noted between those Mk IV implants placed into type 3 bone and those placed into type 4 bone, namely, less energy was required to insert the implants into type 4 bone. Perhaps the most interesting finding was that less energy is used to insert a Mk IV implant than is used to insert a standard implant into each of the three bone types studied.

CONCLUSIONS

Within the limitations of this study, these findings would appear to support the manufacturer's claim that the double thread of the Mk IV implant reduces the insertion time and the energy dissipated to the bone at the bone-implant interface. However, absolute values must be regarded with caution and cannot be used to directly deduce the energy imparted to the bone although if less energy is used during the insertion of a Mk IV implant into the bone, then less energy must be imparted to the bone surface.

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