Dynamic Behavior of Implants as a Measure of Osseointegration

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Research into the formation, destruction, and adaptation of bone around implants would benefit from a sensitive, nondestructive, noninvasive, and quantitative technique to assess the bone-implant interface. It is hypothesized that osseointegration can be quantified by sensing the mechanical impedance (or micromobility) of the implant when it is subjected to minute vibratory forces superimposed upon a quasi-static preload. To test this hypothesis, a total of 24 identical threaded, titanium root-form implants (10×3.75 mm, Osteo-Implant, New Castle, PA) were placed in the mandibles of 4 Walker hounds and allowed to heal submerged for 3 months. The implants were exposed and characterized for osseointegration using clinical observations, quantitative radiography, and a custom-designed impedance instrument. Subsequently, arbitrarily selected implants were ligated to induce bone loss and examined monthly over a 6-month study period. Following the terminal examination and euthanasia, quantitative histologic measurements were made of bone adjacent to the implant, including estimates of both crestal bone height and the percent bone (bone fraction). Linearized dynamic parameters (effective stiffness and effective damping) correlated well with radiographic and histologic measures of bony support (r^2 values ranged from 0.70 to 0.89). Moreover, the presence of nonlinear stiffness was clearly associated with a bimodal "clinical impression" of osseointegration ($P < .0003$, 1-way analysis of variance). These results confirm that, in this animal model, mechanical impedance can be used as a measure of implant osseointegration. (INT J ORAL MAXILLOFAC IMPLANTS 2001;16: 637–645)

Key words: *animal model, dynamics, histomorphometry, impedance, implant, instrumentation, mobility, osseointegration, stiffness*

This report addresses a new technique to observe the state of individual dental implants in situ. In particular, it is hypothesized that quantitative measurements of osseointegration can be accomplished by sensing the mechanical impedance (or micromobility) of the implant when it is subjected to minute vibratory forces superimposed on a quasi-static preload. If validated against other measures of osseointegration, this principle would lend itself to instrumentation that is sensitive, nondestructive, and minimally invasive. Such a capability would be valuable both in fundamental studies of osseointegration kinetics and in the clinical management of individual implants.

The placement of dental implants using the method introduced by Brånemark and coworkers[1] leads to excellent results when performed by experienced operators under optimal conditions. Nevertheless, appreciable numbers of implants fail to integrate normally, leading to loss of the implant and surrounding bone. The risk of failure increases as implants are used in older patients generally, in compromised bone (resulting from osteoporosis or grafting), in the posterior maxilla, in low-density bone, and in the presence of systemic conditions that affect healing, such as diabetes. The growing use of "single-stage" and "immediate" techniques will likely demand more detailed monitoring of individual implants after placement.

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MECHANICAL STABILITY AND OSSEOINTEGRATION

Implant stability occurs in 2 phases: primary stability at placement, and secondary stability as bone healing and remodeling occur at the interface.[2] Despite some reports to the contrary,[3,4] the preponderance of data[5–7] support the common-sense hypothesis that the amount and quality of bone in contact with the implant are important, and mechanical properties are reasonably indicative of the state of osseointegration, especially at the extremes.

THE MEASUREMENT OF OSSEOINTEGRATION

Osseointegration is associated with intimate and long-lasting contact between bone and the alloplastic tooth root replacement material. While there is yet no generally accepted device or method for the objective clinical assessment of osseointegration,[8] techniques used for this purpose include:

- Manual percussion and mobility tests. This is by far the most common clinical technique to assess implants. Typically, successfully functioning implants are immobile and exhibit a clear, ringing sound when percussed, while failing implants tend to be mobile and elicit a dull sound. Admittedly subjective and insensitive to small changes in state, these are nevertheless useful "go/no-go" tests for osseointegration, both initially (before loading) and for periodic follow-up assessment. Unlike histologic alternatives, they are minimally invasive and nondestructive. The use of clinical parameters to evaluate the clinical success and failure of implants is well established.[9]
- Histology. The defining histologic characteristic of osseointegration is the direct apposition of bone to the alloplastic surface with no interposing fibrous tissue at the light microscopic level.[10] This result has been found in humans and in several animal models, although the amount of bone at the interface varies with the animal species and implant type,[4] as well as the site and other factors. No specific amount of bone contact has been adopted as a standard for osseointegration.
- Other destructive techniques. Many other approaches have been suggested to overcome the limits of the qualitative clinical examination: electron microscopic studies, conventional histology, the optical chamber technique, and various biomechanical methods such as torque, push-out, and pull-out testing. Whatever their strengths and weaknesses, all share the disadvantage of being destructive; they are therefore unsuitable for use in human patients and incapable of producing true longitudinal data in any species.
- Radiographic techniques. Radiography, which is noninvasive and widely available, can be used to monitor the physical manifestations of implant osseointegration and failure to approximately 0.5 mm resolution with conventional techniques, or to 0.1 mm with digital subtraction radiography.[11] However, functional osseointegration depends upon phenomena at and within the thin layer of interfacial soft tissue, which cannot be resolved by the usual clinical radiographic techniques,[12] especially on buccal and lingual surfaces masked by the radiopaque implant.
- Mobility and impedance techniques. Several investigators have attempted to quantify the well-known relationship between mobility and implant failure. In 1989, Saratani and associates[13] demonstrated the feasibility of using impulse forces in human subjects. More recently, a device to characterize implant-bone dynamics based on the resonant behavior of a vibrating cantilever, developed by Meredith and coworkers, was used in animal and human studies.[6,14] A commercial mobility instrument[15] (Periotest, Siemens, Bensheim, Germany) has been used to determine implant stability as related to stiffness and damping. All such devices give results that vary with position and angulation and cannot be used until the implant is surgically exposed.

The present study was designed to validate this last category of measurement (mobility—or more generally, dynamic impedance) against 3 other categories (clinical impression, histology, and radiography), using contemporaneous measurements in an animal model. A fifth category, destructive mechanical testing, is incompatible with histology and was not included.

MOBILITY AS AN INDICATOR OF OSSEOINTEGRATION

Most authors cite mobility as a criterion for clinical success, either as an essential, continuous indicator of implant health[16] or as a simple dichotomy that distinguishes success from failure.[17,18] Just as the use of clinical mobility alone cannot always determine the condition of a tooth,[19] neither can it always determine the status of an implant[12]; there are cases of extreme bone loss where there is no apparent mobility.[20] While osseointegration implies a rigid

Fig 1 The impedance handpiece.

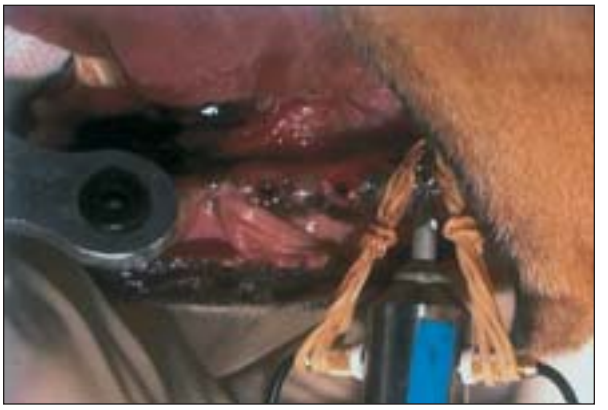

Fig 2 The probe in contact with the implant-abutment assembly, stabilized by an elastic band.

interface with no clinically detectable movement, some studies have demonstrated micromovement of the implant on a subclinical level.[21] Conversely, "fibro-osseous integration" allows a limited amount of clinical implant mobility, a fact recognized in some proposed definitions of clinical success.[22,23]

MATERIALS AND METHODS

The micromobility instrumentation used in this study has been described previously.[7] Briefly, the instrumentation suite consists of a handheld probe, a signal generator/processor unit, and a computer program for data analysis. The probe applies a controlled high-frequency dynamic force roughly perpendicular to the long axis of the implant, superimposes a slowly varying lateral load produced by the examiner's manual pressure, and simultaneously measures force and acceleration wave forms at the driving point as well as the low-frequency preload force wave form.

The handheld probe is shown in Fig 1, and the experimental abutment assembly is shown in Fig 2. The abutment assembly comprised a 4-mm Precision Margin Esthetics transmucosal abutment (Steri-Oss, Nobel Biocare, Yorba Linda, CA) and cast gold healing cap. A small conical indentation was made on the side of the healing cap to permit a stable, reproducible purchase for the probe tip. A single abutment assembly was used in all experiments to eliminate variations in size or mass of the abutment assembly. The abutment assembly was placed on each implant with its conical reference indentation facing buccally. Each component was rigidly attached to the implant with a torque wrench[24] to 10 Ncm; in the case of mobile implants, the implant was grasped with a hemostat before tightening. The operator engaged the healing cap with the probe tip resting in the indentation, pointing approximately normal to the long axes of the mandible and implant (Fig 2). To prevent slippage at low preloads, the probe tip was secured to the healing cap with an elastic band, stabilized by a small wire soldered to the healing cap.

Variations in placement, alignment, and pressure are potential error sources in this and any other handheld instrument. Alignment and pressure errors were effectively eliminated by the reference indentation and by direct force measurements, respectively. Errors in angulation of the handpiece are proportional to $(1 - \cos\theta)$, where θ is the deviation from the normal or other reference vector. If uncontrolled variation in angulation is about ± 10 degrees, it would contribute at most a 1.5% error in impedance magnitude. Because the elastic band acts between the abutment assembly and the impedance head below the load cell, it has no effect on the net measured preload and only an infinitesimal and undetectable effect on impedance.

The effective linearized impedance Z_e of the implant is related directly to the complex ratio of the dynamic force and acceleration signals. A least-squares estimation algorithm generates optimal estimates of the 3 parameters—effective mass (m_{eff}), effective stiffness (k_{eff}), and effective damping (c_{eff})—that completely characterize a second-order linear dynamic model. These parameter values are expressed as functions of preload. If the system is nonlinear, one or more of the parameters will vary with preload; if linear, they will be essentially constant. Previous experiments[7] showed that a second-order linearized model provides an excellent fit to the functional form of the data at a constant preload.

Surgical Protocol

This study used 4 adult Walker hounds. The university animal use committee approved the experimental protocol, and all subject animals were cared for in a humane manner. Extraction of the second, third, and fourth premolars was performed bilaterally in the mandible of each animal. Atropine (0.05 mL/lb) and tiletamine HCl with zolazepam HCl (5 mg/lb) were administered, followed by intubation. A combination of oxygen and isofluorane inhalation was used to induce and maintain each animal under general anesthesia. One gram of cephalexin was administered intravenously. Standard operating room protocol for sterility was maintained at all times. Lidocaine with epinephrine 1:100,000 was administered locally to provide vasoconstriction and analgesia.

Mucoperiosteal flaps were reflected buccally, and releasing incisions were placed lingually. The teeth were segmented cervico-occlusally through the bifurcation with a high-speed dental handpiece and continuous saline lavage and were removed with elevation and forceps extraction. Resorbable gut sutures were placed and finger pressure applied to the wound for 5 minutes. Each animal was given buprenorphine HCl (1 mL) for pain and was observed postoperatively by the veterinary staff. A soft diet was provided, and recovery was uneventful.

After a 3-month healing period, each animal was returned to surgery for implant placement. Again, a continuous mucoperiosteal flap procedure and tissue retraction was performed at each extraction site, and 3 identical implants (10×3.75 mm, Osteo-Implant, New Castle, PA) were placed in each side of the mandible of 4 animals. The implant sites were prepared under continuous saline lavage with sequential enlargement of the crypt size. For this procedure, a Steri-Oss console (model 13555, Nobel Biocare, Yorba Linda, CA) was used with its accompanying handpiece. Drilling and counterboring speeds were both 1,800 rpm. Approximately 5 mm was allowed between implant sites. Threading of the implant site was performed at less than 50 rpm. Each animal received 6 implants optimally placed, with the exception of 1 implant, which was deliberately plated with elemental tin (Sn) to prevent osseointegration. During the recovery period, the veterinary staff performed follow-up care.

The implant sites healed for 3 months. Recovery and wound healing were uneventful. At the end of 3 months, each animal was returned to surgery and the implants were exposed.

Clinical and Laboratory Measurements

Each implant was evaluated monthly during a 6-month period. Standardized radiographs, aligned with preformed film holders, were made at each examination to permit quantitative subtraction. The abutment/healing cap assembly was positioned and secured as described. At each examination, each implant was first tested for clinical osseointegration by a single examiner using the conventional method of tapping with a metallic instrument (percussion testing). Additionally, each implant was examined for the presence of clinical mobility by pushing the implant back and forth between 2 blunt metallic instruments. Any implant with either palpable mobility or dullness to percussion was classified as "clinically nonintegrated." Clinically immobile implants with a bright, ringing sound were deemed "clinically integrated." These measurements were performed on each implant in isolation without reference to adjacent implants.

The experimental instrument was then used to characterize each implant. The probe tip was placed on the implant-abutment assembly to exert a preload normal to the corono-apical axis of the implant. The vibrator was then actuated to produce a burst of ten 750-Hz half-sine pulses separated by approximately 100 ms. The operator randomly varied the preload on the implant in the range 0 to 5 N by means of gentle hand manipulation of the instrument. Twenty coordinated time histories (measurements) of oscillatory force, acceleration, and preload data were acquired, digitized, and saved for subsequent analysis. The procedure was repeated for each implant.

Baseline data for each measured parameter were acquired at the initial monthly examination. At the end of the second monthly examination, implants were assigned at random to 3 groups: 6 were left to integrate naturally, 17 were fitted with cotton ligatures to induce bone loss, and the single tin-plated implant (which had failed by this time) was left in place. The purpose of this intervention was to produce a wider range of bone status than would be expected to occur naturally. Although ligatures induced significant bone loss, in no case was this sufficient to cause outright failure before the terminal appointment. Therefore, at the fifth monthly examination, arbitrarily selected implants were subluxated using extraction forceps until palpably mobile and left in place.

The identical sequence of measurements was repeated at the sixth (final) appointment, after which the animals were euthanized and the implants retrieved. Of the 24 implants placed, a total of 7 were found to have clinically failed from the following causes: natural failure to integrate (n = 1), failure to achieve osseointegration or loss of osseointegration from natural causes (n = 1), and induced failure by ligatures and subluxation combined (n = 5).

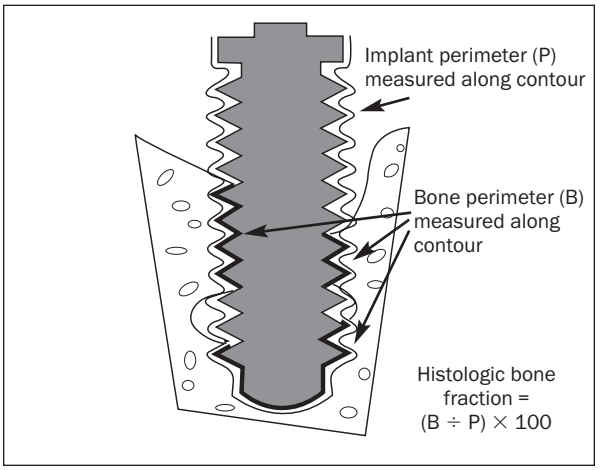

Fig 3a Bone length apposing the implant is measured relative to the implant surface length.

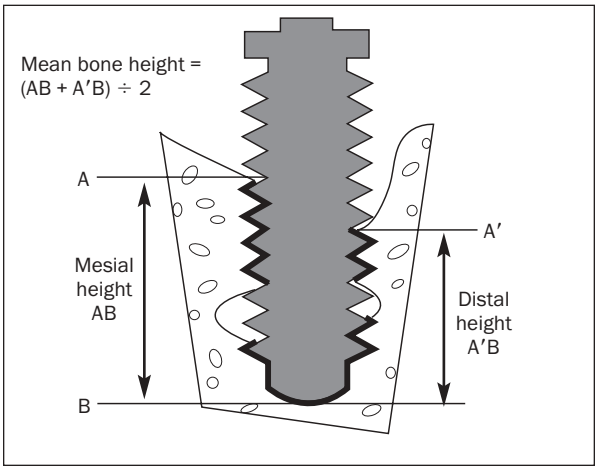

Fig 3b Crestal bone height is measured radiographically and histologically from the apex of the implant to the coronal terminus.

Histologic Examination

The retrieved mandibles were placed in 10% neutral buffered formalin (Surgipath Medical Industries, Richmond, IL) for 48 hours. Each mandible was sectioned mediolaterally between adjacent implants with a rotary disk, and the blocks were stored in 10% formalin solution. The specimens were dried in graded alcohol solutions and embedded in clear polymethyl methacrylate resin under vacuum. The embedded implants were ground along the mediolateral plane of the mandible to obtain longitudinal sections at the 25% level along the implant diameter. The specimens were stained with van Gieson's solution (Polyscientific Research Corporation, Bay Shore, NY).

All specimens were examined using reflected light microscopy at 135×. The fraction of bone in contact with the implant at the interface (expressed as a percentage of the perimeter of the implant section) and the height of the bone projected on the implant long axis (representing the mean of buccal and lingual measurements) were measured using a computerized histomorphometry system (Micro-Vu Corporation, Windsor, CA) (Figs 3a and 3b). Twenty-four values of bone fraction and bone height were thus determined. No attempt was made to differentiate the type of bone in contact (ie, cortical or trabecular), and the resolution of this technique does not permit measurement of proteoglycan thickness. While the amount of bone in contact with the implants varied widely, none of the implants engaged the lower cortical plate of the mandible.

Radiographic Examination

Using the method described by Jeffcoat and coworkers,[25] the authors performed quantitative measurements of crestal bone height, as measured from the apical aspect of each implant on radiographs taken at each interval. An average crestal bone height was calculated from the mean of mesial and distal measurements for each of the 24 implants (Fig 3b).

Dynamic Postprocessing

Optimal estimates of the dynamic parameters were obtained individually for each of the 5,760 test ensembles (20 measurements × 24 implants × 2 repetitions × 6 examinations). This was accomplished by comparing the measured driving-point acceleration with that of an ideal second-order model driven by the measured dynamic force. Parameter values and initial conditions were systematically adjusted to reduce the integral square error between the observed and simulated responses, until a local minimum was reached. The resulting set of 3 dynamic parameters (k_{eff}, c_{eff}, and m_{eff}) was augmented with the mean quasi-static preload (F_o) measured over the sampling period. The process was entirely automated to eliminate the possibility of bias or variability on the part of the analyst.

According to the working hypothesis, a particular type of nonlinearity, in which stiffness increases with preload, would characterize absence of osseointegration. As a consistent measure of this nonlinearity, a "hardening coefficient" (HC_Q) was defined as the derivative of any normalized parameter Q (eg, k_{eff} or c_{eff}) with respect to preload, $\delta Q/\delta F_o$. HC_Q was calculated using the slope and intercept of a linear regression line fitted to 20 values of Q computed at various preloads. The computed slope was finally normalized by the estimated value of Q at 0 preload (intercept). A value of HC_Q near 0 corresponds to a nearly linear system, while a stiffening nonlinearity

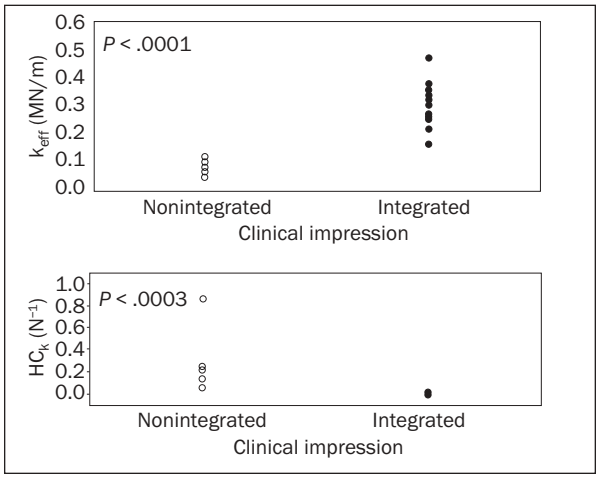

Fig 4 Effective stiffness k_{eff} (upper curve) and nonlinear hardening coefficient HC_k (lower curve) versus clinical impression.

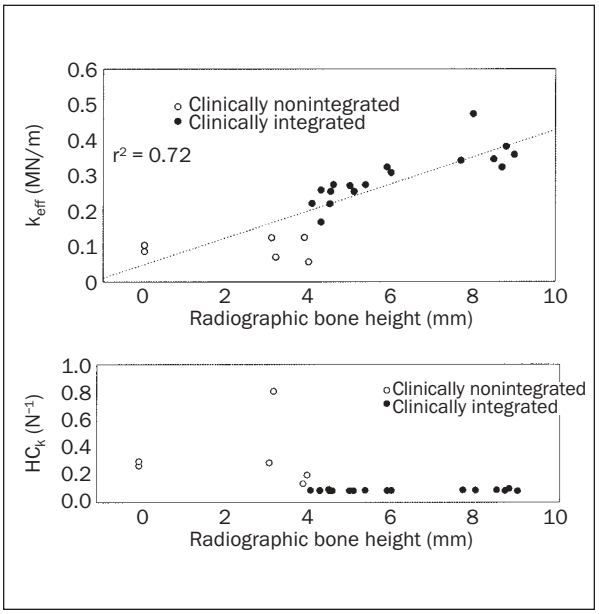

Fig 5 Effective stiffness k_{eff} (upper curve) and nonlinear hardening coefficient HC_k (lower curve) versus radiographic bone height.

will exhibit a significant positive hardening coefficient. The HC for k_{eff} and c_{eff}, as well as mean values of k_{eff}, c_{eff}, and m_{eff} over all preloads, were computed for each implant individually at each examination. At each appointment, a total of 960 test ensembles were examined, yielding 2 values for k_{eff}, c_{eff}, HC_k, and HC_c for each of 24 implants. (Implant mass parameters, m_{eff} and HC_m, were also computed in the same way. Estimated mass was linear and differed only slightly from the known static mass. While this fact provided a valuable confirmation of the method's validity, it also justified exclusion of mass as a diagnostic parameter.) After a 2-way analysis of variance (ANOVA) confirmed that there were no significant differences between repeated measurements, the 2 parameter estimates were averaged and the mean used for all subsequent analyses. In summary, the data recorded for each implant at each examination consisted of:

- Linear dynamic parameters, k_{eff} and c_{eff}
- Nonlinear dynamic parameters, HC_k and HC_c
- Clinical impression (binary)
- Radiographic bone height (mm)
- Histologic bone height (mm)—terminal examination only
- Histologic bone fraction (%)—terminal examination only

A single examiner performed all measurements.

RESULTS

This article compares clinical, histologic, radiographic, and dynamic characterizations of osseointegration at the terminal examination; the kinetics of bone gain and loss over the 6-month experiment will be reported separately. For the statistical analysis, k_{eff} and HC_k were compared separately against 4 independent variables: clinical impression, radiographic bone height, histologic bone height, and bone fraction. An equivalent analysis using c_{eff} and HC_c demonstrated similar trends in every case, albeit somewhat less pronounced.

For easy comparison, Figs 4 to 7 present the main findings in a common format, where the horizontal axis is one of the 4 reference measures. The vertical axes represent the linear (k_{eff}, upper curve) and nonlinear (HC_k, lower curve) dynamic parameters. Open circles refer to clinically nonintegrated implants and closed circles to integrated implants. Correlation and measures of significance are shown on the figures where appropriate.

Correlation Between Dynamics and Clinical Impression

Figure 4 illustrates the correlation between clinical impression and the dynamic stiffness parameters evaluated in this study. Each implant was categorized as "integrated" or "nonintegrated," with the presence of either a dull sound on percussion or

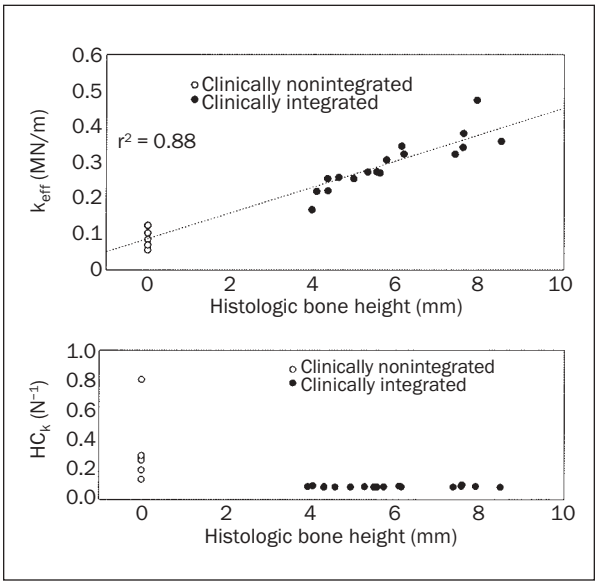

Fig 6 Effective stiffness k_{eff} (upper curve) and nonlinear hardening coefficient HC_k (lower curve) versus histologic bone height.

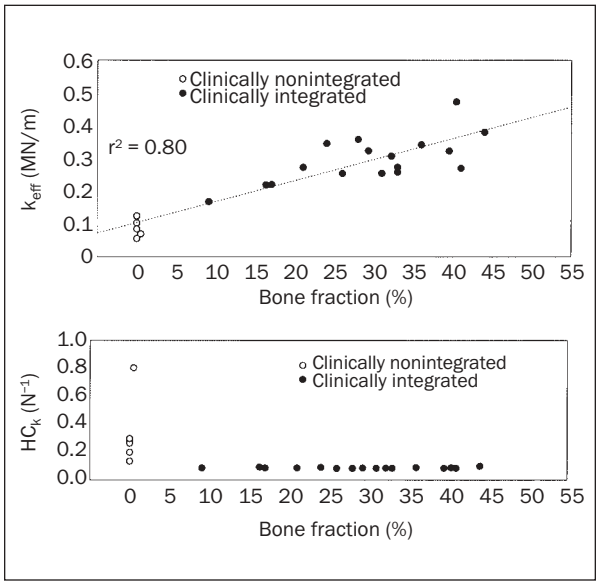

Fig 7 Effective stiffness k_{eff} (upper curve) and nonlinear hardening coefficient HC_k (lower curve) versus bone fraction.

perceptible mobility being sufficient to classify an implant as nonintegrated. These clinical characterizations from the terminal visit were used as the independent factor in a 1-way ANOVA, in which the dependent variable, HC_k, is a measure of the nonlinearity of k_{eff}. Such nonlinear behavior was found to be pronounced in clinically failed implants. Results showed a highly significant difference ($P < .0003$, n = 24) in HC_k between clinically integrated (n = 17, $\mu = 0.005\ N^{-1}$, $\sigma = 0.005\ N^{-1}$) and clinically nonintegrated implants (n = 7, $\mu = 0.30\ N^{-1}$, $\sigma = 0.29\ N^{-1}$).

The mean value of k_{eff} in clinically integrated implants was 0.29 MN/m, ranging from 0.16 to 0.49 MN/m. In clinically failed implants, the mean value of k_{eff} was 0.096 MN/m, ranging from 0.053 to 0.15 MN/m. This difference is significant at the $P < .0001$ level.

Correlation Between Dynamics and Bone Height

The level of bone (crestal bone height in apposition to the implant longitudinal cross-section) was measured radiographically at all examinations, and histologically at the terminal visit. The 2 methods agree reasonably well with one another ($P = .01$, dependent paired t test, n = 24). However, radiography is the less sensitive method for detecting nonintegration, as is evident from inspection of Figs 5 and 6: The clinically nonintegrated implants were found to have zero histologic bone contact but varying amounts of apparent radiographic bone support. This result is not surprising, since while conventional radiography provides a good measurement of gross bone height, it cannot resolve fine structures immediately adjacent to the implant structure.

Both methods show increasing values of k_{eff} as a function of bone height, although as expected, the correlation is higher for histologic ($r^2 = 0.88$) than for radiographic ($r^2 = 0.72$) measurements. The linearity of the integrated implants was unaffected by bone height (Figs 5 and 6, lower curves).

Correlation of Dynamics with Bone Fraction

Effective stiffness and damping, measured in vivo at the terminal examination, were both found to correlate positively with the amount of bone in contact with the implant as measured in histologic samples. Figure 7, in which the independent variable is bone fraction and the dependent variable is k_{eff}, is representative. The upper curve plots k_{eff} versus bone fraction; the lower shows the stiffness hardening coefficient HC_k versus the same quantity.

Examination of Fig 7 leads to 2 conclusions, both of which support the hypothesis. First, k_{eff} increases in a roughly linear fashion ($r^2 = 0.80$) with the fraction of bone apposing the implant surface. Second, a clinical impression of integration observed over a wide range of (nonzero) bone fraction is clearly associated with the absence of a hardening nonlinearity

in the dynamic response (see the ANOVA results presented above).

The wide variation in dynamic properties of nonintegrated implants, a striking feature of Fig 7, corresponds to an intuitive distinction between "flexible" and "loose" support. An implant supported by bone, even to a small degree, returns promptly (if weakly) to its original position when an external force is removed, whereas the response of a nonintegrated implant is determined by the vagaries of its encapsulation.

DISCUSSION

This study examined the dynamic response of root-form implants under a variety of simulated clinical conditions, including: clinically osseointegrated with minimal bone loss, clinically osseointegrated with variations in the amount of bone loss, failure to osseointegrate, and loss of osseointegration resulting from progressive bone loss or imposed trauma. Measured quantities included k_{eff}, c_{eff}, and HC_Q (the nonlinear change in k_{eff} or c_{eff} caused by an applied load), along with more customary measures of osseointegration.

The essential results are twofold. First, clinically nonintegrated implants exhibit a nonlinear stiffness when subjected to an external bias load, whereas integrated implants show no such behavior. Second, among integrated implants, the value of dynamic stiffness is roughly proportional to the amount of bone in contact with the implant. Similar findings apply (though at lower significance levels) to the dynamic damping coefficient. While the data reported here are necessarily limited to a single terminal examination, at which histologic samples were available, intermediate monthly examinations showed the same relationships between dynamics and the other available measures.

One may ask whether the observed dynamic parameters are attributable to the implant-bone interface or to some other source such as the surrounding bone, the implant geometry, or unmodeled flexibility of the implant-abutment system (see Meredith[26]). In a previous study using dog tibiae,[7] Ramp and coworkers demonstrated that the instrument caused implant displacement of about 15 nm, well within the limits of the thickness of the proteoglycan layer typically found at the implant-bone interface. Moreover, the present experiment was designed to eliminate, insofar as possible, all variation in geometry and structure (identical implants, common abutment, and uniform surgical technique). These facts, together with the wide but consistent variation in parameters among otherwise identical implants, suggests that the stiffness being measured is dominated by that of the immediate interface, not the surrounding tissue or flexibility of the implant. The results, of course, would be quantitatively different if a different implant/abutment system had been used.

Interpretation of measured implant response, in isolation, is problematic. The absolute value of impedance, no matter how accurately measured, may not be sufficient for diagnostic purposes, because at least 4 other important sources of variation are uncontrollable or difficult to compensate: *(1)* anatomic placement, *(2)* bone condition (and the spatial distribution thereof), *(3)* axis and point of loading, and *(4)* conditions of loading (particularly preload and amplitude). Nevertheless, with the acquisition of baseline data at the time of placement, trends in effective stiffness and damping may be useful for the longitudinal observation of implants from placement to exposure and restoration. In the absence of such baseline data, these factors complicate interpretation of implant response. Appropriate nonlinear parameters can help overcome these difficulties, if they are substantially self-normalizing.

The high degree of concordance with histologic, clinical, and radiographic assessment warrants continued investigations into impedance methodologies for the characterization of osseointegration. Linear parameters appear to be useful surrogates for total bone fraction and crestal bone height, both of which are primary indicators of implant stability. A pronounced (nonlinear) variation in k_{eff} with loading provides an objective and unambiguous delineation of the nonintegrated implants.

SUMMARY

This research provides 4 main findings that may be of value in future studies of osseointegration.

- First, the dynamic parameters k_{eff} and c_{eff} correlate well with clinical and histologic assessments of osseointegration.
- Second, these parameters are stable (over the time period investigated) in the integrated implant, and so provide a basis for longitudinal examination of osseointegration.
- Third, nonlinear characteristics of implant mobility may be useful in the objective determination of osseointegration.
- Fourth, useful clinical instrumentation for the objective evaluation of osseointegration can be

based on impedance principles, particularly given a repeatable, noninvasive probe configuration for use during the postplacement healing and remodeling phase.

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