Volume 6: Clinical Performance and Enhanced Stability
February 2008

ISSN 1651-0070
We are all concerned with the clinical safety of implant systems, be they dental implants, maxillofacial systems or orthopaedic components. In particular, the means by which an implant system might be proven safe before release for clinical use or even marketing continues to attract debate or criticism.

Computer modelling and laboratory testing are two well tried methods which can give useful indications in the search for safety of implants. Laboratory tests under accurately known and measured conditions, with carefully controlled variables and environments, can be helpful in comparing design solutions or competitive systems for similar applications. But systems which interface living bone and function within unpredictable human subjects can be subject to many factors for which we have little data and can provide insufficient predictions. Under such constraints, laboratory tests may not be realistic, and may give unreliable indications of implant performance in clinical practice (unreliably pessimistic or unreliably optimistic). In all implant fields one can recall stories of systems launched with claims of laboratory validation, and complex and detailed test programmes in support of their clinical performance, which nevertheless demonstrated poor relevance to clinical realities and ultimately gave poor results in clinical practice.

In the end, good clinical practice and good engineering lead to the same conclusion – the ultimate test or proof is in the real situation; the patient. Laboratory and computer testing and modelling may or may not give confidence that every risk has been accounted for and every aspect of predicted service is accommodated in the system design and production. The real test (the “gold standard”) is performance in the clinical setting. Good engineering and thorough pre-clinical testing will reduce the risk to patients and the number of “clinical trial patients” required to be exposed to unproven components or systems, but no amount of careful pre-clinical testing can remove the need for carefully designed and monitored clinical trials.

So clinical results are the only complete and reliable proof of a system’s suitability for clinical use.

Warren Macdonald
Visiting Lecturer, Department of Biomaterials, Institute of Clinical Sciences, University of Göteborg, Göteborg
SWEDEN
The Neoss Implant System

Treatment with osseointegrated implants is an integral part of contemporary patient care. Many companies provide the dental community with the necessary products. Quite frequently they also provide us unnecessary products! What should I use? Which implant system is the best? Naturally, there is no single or easy answer to such a question. It’s a matter of preferences. These preferences will of course vary from individual to individual, but also reflect different needs of the surgeon, the prosthodontist and the dental technician. Not to forget the staff member responsible for the inventory of the clinic!

The founders of the Neoss company, Dr. Neil Meredith and Engineer Fredrik Engman, have created an implant system that should fulfil all needs of versatility, flexibility and ease-of-use, but through a minimum of components and instruments.

I admit to finding the thinking and the design features of the Neoss implant system attractive. But does it work? What’s the scientific evidence? In this issue of Applied Osseointegration Research you will find a mix of clinical studies, animal experiments and laboratory tests evaluating the Neoss implant system.

The osseointegration process is influenced by the mechanical stability of the implant at surgery and the following biological response. The mechanical stability is related to the macro-design of the implant whereas the surface properties will influence the biological response. The modern moderately rough surfaces have demonstrated strong bone responses, allegedly shortening the healing time. However, the Neoss is a modern implant system that is, actually minimally rough (as demonstrated in surface topographical studies). Despite this the Neoss system obviously achieves quite positive results. Still, when using immediate or early loading protocols, the system is challenged by loading the bone simultaneously with the healing process and development of stability.

The papers accepted for publication in this issue have assessed the mechanical stability of Neoss implants at surgery by insertion torque values and ISQ values (resonance frequency analysis). The biological response during healing is documented using histological analyses and biomechanical tests such as removal torque and ISQ values. The clinical outcome of two-stage, one-stage and immediate loading protocols has been evaluated at one-year follow-up by clinical and radiographic examinations. The biocompatibility of PEEK healing abutments is evaluated by means of bacterial colonization and real-time PCR (polymerase chain reaction) tests. The outcomes are, in my mind, excellent. Enjoy!

Jan Gottlow, DDS, Ph D
Guest-Editor-in-Chief of Volume 6
Department of Biomaterials,
Institute of Clinical Sciences,
University of Göteborg,
Göteborg
SWEDEN
# Table of Contents

A REVIEW OF IMPLANT DESIGN, GEOMETRY AND PLACEMENT  
Neil Meredith  
6

HISTOLOGICAL EVALUATION OF A BIMODAL TITANIUM IMPLANT SURFACE. A PILOT STUDY IN THE DOG MANDIBLE.  
Luiz A Salata¹, Paulo EP Faria¹, Marconi G Tavares¹, Neil Meredith²,³ and Lars Sennerby⁴  
13

HISTOLOGICAL AND BIOMECHANICAL ASPECTS OF SURFACE TOPOGRAPHY AND GEOMETRY OF NEOSS IMPLANTS. A STUDY IN RABBITS.  
Lars Sennerby¹, Jan Gottlow¹, Fredrik Engman¹ and Neil Meredith²,³  
18

A ONE-YEAR CLINICAL, RADIOGRAPHIC AND RFA STUDY OF NEOSS IMPLANTS USED IN TWO-STAGE PROCEDURES  
Peter Andersson¹, Damiano Verrocchi¹, Rauno Viinamäki¹, Lars Sennerby¹,²  
23

IMMEDIATE/EARLY LOADING OF NEOSS IMPLANTS. PRELIMINARY RESULTS FROM AN ONGOING STUDY  
Peter Andersson¹, Damiano Verrocchi¹, Luca Pagliani², Lars Sennerby³  
27

A RETROSPECTIVE FOLLOW-UP OF 50 CONSECUTIVE PATIENTS TREATED WITH NEOSS IMPLANTS WITH OR WITHOUT AN ADJUNCTIVE GBR-PROCEDURE  
Thomas Zumstein¹ and Camilla Billström²  
31

INSERTION TORQUE MEASUREMENTS DURING PLACEMENT OF NEOSS IMPLANTS  
Luca Pagliani¹ Lars Sennerby² Peter Andersson³ Damiano Verrocchi³, Neil Meredith⁴  
36

STRESS EVALUATION OF DENTAL IMPLANT WALL THICKNESS USING NUMERICAL TECHNIQUES  
Rudi C. van Staden¹, Hong Guan¹, Yew-Chaye Loo¹, Newell W. Johnson³, Neil Meredith¹  
39

COMPARATIVE ANALYSIS OF TWO IMPLANT-CROWN CONNECTION SYSTEMS - A FINITE ELEMENT STUDY  
Rudi C. van Staden¹, Hong Guan¹, Yew-Chaye Loo¹, Newell W. Johnson³, Neil Meredith³  
48

COMPARISON OF EARLY BACTERIAL COLONIZATION OF PEEK AND TITANIUM HEALING ABUTMENTS USING REAL-TIME PCR  
Stefano Volpe⁴, Damiano Verrocchi², Peter Andersson¹, Jan Gottlow¹, Lars Sennerby²,³  
54

SURVIVAL RATE, FRACTURE RESISTANCE AND MODE OF FAILURE OF TITANIUM IMPLANTS IN CLINICAL FUNCTION AND DYNAMIC LOADING.  
Neil Meredith¹,² and Fredrik Engman²  
57
A Review of Implant Design, Geometry and Placement

Neil Meredith
University of Bristol, UK

INTRODUCTION
Bone is a unique structural material. Its physical and mechanical properties mimic both natural materials such as wood and man-made materials including polymers. Its mechanical properties can be directly correlated to its complex structure and it should therefore be described as an anisotropic material (Carter & Beaupre, 2001). In contrast many man-made polymers have a uniform structure in all directions and can thus be considered homogenous materials. Bone’s unique property is its ability to form new bone and to remodel existing bone. This is especially important in its response to applied mechanical stresses (Carter et al, 1998).

Bone healing considerations
In the complete absence of stress bone can and will form and remodel; as with the bone formation occurring during osseointegration in the submerged phase of a two-stage implant (Lekholm & Zarb, 1985). There is also evidence to show that the application of light mechanical loads can induce favourable stresses within the bone, which may induce accelerated and enhanced bone formation (Pilliar & Maniatisopoulos, 1986) Much research has been carried out to analyse the influence of mechanical stress on bone formation, but it is extremely difficult to model these parameters in a laboratory or even an animal model (Brunski, 1999, Brunski et al, 1993).

It is clear that the application to bone of mechanical stresses in excess of a certain threshold level will not induce bone formation but can lead to the formation of a fibrous tissue capsule and malunion, or in the case of dental implantation, a failure of osseointegration (Szumkler-Moncler et al. 1998). This threshold is not clearly known or defined and is likely to vary in individual cases and sites in relation to bone quality and quantity, the loading conditions applied, and the systemic and regenerative capacities of the patient (Brunski et al, 2001). There is clearly a need to optimise the factors associated with implant placement and design which can optimise the conditions for bone remodelling or formation under immediate and early loading conditions (Sennerby & Roos, 1998).

Biomechanics of implant placement and loading
There are three main biomechanical parameters that influence the stress distribution and optimal stability of an implant in bone; these are the placement procedures including the drilling of the osteotomy site and the use of compression techniques to increase local stability (O’Sullivan et al. 2004a). Secondly, the design features of the implant itself (O’Sullivan et al, 2000) and thirdly, the loading conditions to which the implant is subjected (Friberg et al, 1991).

Loading conditions may differ due to a single or two-stage surgical placement technique (Sennerby & Roos, 1998). In a two-stage technique, the implant fixture is placed, typically level with the crest of the bone and submerged beneath the soft tissue, for a healing period which may vary but has historically been recommended as three months in the mandible and six months in the maxilla. A two-stage protocol effectively eliminates any dynamic functional loads during this healing and osseointegration period. A single-stage technique is different in that the implant is exposed to the oral environment at the time of initial surgery and placement. In this case, a number of options exist in that the implant may be loaded (immediate loading) or may remain unloaded by the provision of a relieved prosthesis over the implant site for a healing period, which may typically vary from six to eight weeks.

Loading protocols
The alternative is early or immediate loading, in which a prosthesis or temporary restoration is placed directly at the time of surgery or a short period thereafter; perhaps a week. Immediate loading and delayed one-stage loading are very different. In delayed one-stage loading it is likely that the effectively unloaded implant is subjected to small dynamic loads, applied through the soft tissue and through intermittent contact with a prosthesis (Orenstein et al.; 1998). The clinical evidence for this protocol is that it is highly successful and that the dynamic loads if any are small enough and of appropriate frequency that they do not lead to a failure of osseointegration and
formation of a fibrous tissue capsule. However, there is no strong evidence to suggest that this early loading one-stage technique will actually accelerate bone formation or enhance the quality of bone formed.

Immediate and early loading, attachment of a prosthesis at the time of implant placement or shortly, thereafter relies on the principal that the dynamic functional loading applied will be below the threshold which can induce failure of osseointegration and formation of a fibrous tissue capsule, and at a level whereby bone remodelling and formation may progress unhindered or even accelerated in the early healing stage. The ranges of clinical, anatomical and surgical parameters are very wide and therefore at present the selection, use and success of immediate and early-loading techniques are generally founded on the experience, knowledge and understanding of the clinician on an individual case by case basis.

Primary Implant Stability
It is clear that one of the keys to successful osseointegration is the primary stability of an implant. It is considered highly desirable that this level of stability should be as high as possible. In experienced hands it is commonly measured by the insertion torque necessary to place the implant. This is clearly a subjective feeling, but can give an experienced operator a level of confidence in the prospects of a successful outcome for a case.

Electronic drill controllers displaying graphs of insertion torque are available, but the interpretation of such data is difficult (Friberg B, Sennerby L, Roos J; 1995). An alternative technique is resonance frequency analysis (RFA)(Meredith N, Cawley P, Alleyne D; 1996), which has been available for some ten years. It utilises a non-destructive test method to measure the local interfacial stiffness of an implant and surrounding bone.

Secondary Implant Stability
It is evident in the first six to eight weeks following placement that there is more new bone formation in poorer quality bone, typically in the anterior maxilla. Here the blood supply can be good but there is an open trabecular network and primary stability is typically lower than in the mandible for example. In the mandible, there are smaller changes in stability and these may be accompanied by local remodelling rather than new bone formation (Meredith et al. 1997). This can result in a measurable increase in stability in bone qualities 3 and 4, more marked than in bone quality 1 and 2 (Andersson et al. 2007).

Clinical findings indicate that a large part of the healing and remodelling process, which is termed osseointegration, is probably completed within the first eight weeks of placement, under normal conditions. Higher primary stability at placement is also measured (Meredith N, Bok, K, Friberg, Sennerby; 1997) in better quality and denser bone. O’Sullivan (2001) measured implant stability as a function of the changes in strain following implant placement for a period of two hours following placement. He observed a sharp initial fall in stability and decrease in interfacial strain.

This is interesting because it is not a biological or physiological phenomenon, what is occurring is mechanical stress relaxation in the bone following placement. This suggests that although a high level of stability and compression may occur at the time of insertion, this stability may decrease very rapidly thereby creating a higher risk situation. This could be especially important where implants are placed in poor bone quality, or where the bone has been artificially compressed to a high level by the use of osteotomes for example.

It is clear therefore that there is a very complex and subtle inter-relationship between bone quality, stability, geometry and placement technique.

Implant Geometry
A range of geometries has been available for dental implants for a number of years and their variation and relation to success do warrant some observations. It is interesting to note that historically the cylindrical implant has been associated with a relatively high in-
Incidence of implant failure (Albrektsson T, Sennerby L; 1991). This has also born some relationship to specific systems and it is important to be certain that the aetiology of this failure is due to the geometry alone; the evidence suggests that this geometry could play a part. However, by relatively small modifications in geometry and placement technique it is possible to achieve a highly successful implant system, the Straumann (Basle, Switzerland) system was essentially cylindrical but with a small widely spaced thread super-imposed. The drilling protocol for this implant uses a small modification with the implant being 0.05mm larger than the preparation site. The idea is that this creates localised compression around the implant. In general, this seems to be successful, although the screw pattern used is that of a Thorpe screw, which was designed for use in orthopaedic surgery to pull bone plates together (Ansell R, Scales J; 1968).

Threaded implants with a close pitch and a deep profile (Brånemark; Nobelbiocare, Gothenburg, Sweden) are quite typical of another design of implant which has been clinically highly successful. The variations in stress distribution between different implant systems under applied loading would therefore appear to be quite substantial. However, both these geometry types (the threaded cylinder and the threaded implant) are equally successful clinically. Implant geometries have not been routinely designed for use in specific bone or quality types. However, implants having a slightly tapered geometry (approximately 4° from parallel) have been introduced to create compression in poorer bone qualities and optimise stability (Brånemark MkIV; Nobelbiocare, Gothenburg, Sweden). Figure 2 illustrates the stability at placement for a number of implant types in the human cadaver maxilla (O’Sullivan, Sennerby, Meredith; 2000).

Static and dynamic stresses

What is apparent is that two separate issues need to be considered in the successful placement and loading of a dental implant. These are the dynamic stresses experienced by an implant in function and the static stresses encountered during implant placement. Dynamic stresses and applied loads can be considerable once an implant has reached an equilibrium position in bone and healing has taken place. Prior to this the nature, magnitude and direction of applied dynamic loads can influence the treatment outcome (Goodman et al. 1993). Static loads at the time of implant placement, however, are a consequence of those techniques or geometry that are designed to contribute to the maximal stability of an implant in bone. The use of slightly tapered threaded implants is one such way of seeking to increase stability in poor bone qualities (O’Sullivan 2001). Is compression and the application of high levels of primary stress a clinical issue? The answer is potentially yes, both at the time of placement in high initial bone qualities and at the time of abutment connection and...
loading in two-stage implants (Misch C E, 1990). A number of implant systems use a drilling sequence that creates a preparation site that is only very slightly smaller than the implant, being inserted. This will quite obviously create quite low levels of insertion stress. The evidence is that these implants are highly successful in average to good bone qualities where compression and stability are adequate and over-compression is avoided. In systems inducing a high level of compression, possibly by use of a tapered implant or a drilling sequence with a drill size much smaller than the implant diameter there are occasions when the consequences of over-compression may be early failure of the implant itself (Hobkirk JA, Rusiniak K, 1977) (Ivanoff C-J 1999).

Figure 3. illustrates the variation in insertion torque for 3.75 mm implants placed in final osteotomies of differing diameter. This clearly demonstrates a relationship between compression of the osteotomy site and insertion torque. (O’Sullivan, 2001)

Most implant designs do not directly address the optimisation of stability in poor bone qualities. Some implants have become available with a slightly tapered geometry, which creates local compression and thereby achieves good stability; these have been very successful. The advantage of having a geometrical feature on the implant is that it does not rely on a complex placement or drilling protocol to create a level of compression, which may be variable in relation to its outcome. The slightly tapered implant thus provides a simple way of repeatedly and consistently offering an increase in stability in poor bone qualities (O’Sullivan et al., 2004b).

A possible thread refinement from current implants is altering the implant to bone volume ratio within the threads by reducing the thickness of the implant threads. An important feature of the geometry of the thread cutting face is that there is adequate volume in the relief chambers for bone clearance. Many of today’s implant systems use a thread-cutting geometry to tap a thread into the bone during insertion into a cylindrical hole. This works very effectively and creates a high level of bone-to-implant contact.

It is therefore desirable to have an implant with a positive tolerance (taper) which will cause optimal compression in poor bone qualities. In order to address the seating, stability and compression levels within good bone qualities further consideration needs to be paid to the surface geometry. Many of today’s implant systems use a thread-cutting geometry to tap a thread into the bone during insertion into a cylindrical hole. This works very effectively and creates a high level of bone-to-implant contact.

Figure 3. Variation in insertion torque with drilling depth as a function of time for 3.75mm implants placed in final osteotomy diameters 2.85-3.7mm. (O’Sullivan, 2001)

Figure 4. Bone relief chambers and cutting face
mal bone to implant contact on the threaded area. Historically some implant designs have been less successful because the bone collection chambers were very wide and very openly spaced (Friberg et al. 1997). It is also important that the cutting face of the self-tapping implant feature is sharp and without burrs. In order to optimise the placement of a threaded implant, having a positive tolerance in good bone qualities, a secondary cutting feature can be introduced along the side of the implant (Figure 4 and 5). This secondary cutting face is much shallower than the apical cutting face and actually does not engage in soft bone qualities. In dense bone however, when the implant is inserted there will be elastic recovery during the insertion process such that the bone will engage the secondary cutting faces and a small amount will be removed.

Figure 6 illustrates an insertion torque profile for the Neoss implant demonstrating a near linear increase in insertion torque with a modulated plot attributable to secondary cutting (Luca et al., 2007). This will not impair the stability of the implant in the site but will create a different fit in dense bone quality or weak bone, thereby optimising stability in all bone qualities but without over compressing good bone. The combination of features relating implant geometry with forming and thread cutting in bone of different quality has been combined into a single implant design and the concept is called TCF which represent a Thread Cutting and a thread Forming implant leading to cutting for optimal seating and forming for optimal stability.

Screw taps work differently from the use of progressively increasing drill sizes to match implant diameters. They are available for use on rare occasions where there is uniformly dense cortical bone along the whole implant length. In such cases, the compression occurring at the implant tissue interface is different from that under typical conditions where there is a combination of cortical and trabecular bone.

Under normal conditions the TCF feature of the Neo implant is designed to create an optimal level of compression, starting from the apex of the threads as the implant is inserted into a cylindrically prepared site. In uniformly dense bone, the use of a screw tap to pre-tap the site will create a level of compression on implant insertion that is uniformly applied to both the peaks and troughs of the thread, thereby achieving a comparable overall level of compression in a structurally different quality of bone. Screw-taps are therefore recommended on occasions where bone density is extremely high and bone quality is very homogenous.

**Insertion Torque**

Insertion torque is a commonly assessed qualitative parameter during implant placement. Placements and procedures vary considerably between systems and between operators. Some clinicians favour a very high level of final insertion torque and other operators work with a very gentle insertion technique, rarely encountering an insertion torque greater than 30Ncm.

Historically, a high final peak insertion torque may be variously affected by three phenomena. The implant may bottom out in the site, so that the final tightening torque is actually inducing very high levels of shear stresses at the implant tissue interface, as the implant butts against the bottom of the preparation site. A second cause of a high insertion torque may be contact at the flange of the implant with the crestal cortical plate, thereby achieving a high level of compression and static stresses. The third reason is a very high level of interfacial stresses leading to a high level of implant stability. On occasion, this may or may not accompany a level of over-compression and very rarely it may lead to implant failure.

The Neoss implant system has been designed to achieve the optimal level of compression and stability, for implant placements of bone in all qualities. This does not rely on a very high level of final insertion torque. Excellent results can be obtained using a gentle technique where the insertion torque at any period dur-
The optimal tissue response can be expected to be achieved if an insertion torque between 25 and 30Ncm is obtained. Historically, if implants fail to seat or encounter very high insertion torques, surgeons typically remove the implant and use a screw-tap to aid the preparation. In the case of a Neo implant, if the insertion torque during placement reaches high levels then the implant can simply be anti-rotated a few turns and then re-inserted. This clears bone swarf from the cutting faces of the implant and reduces friction at the interface, thereby allowing smooth insertion without any risk of over-heating.

A second feature that is important during implant placement is the static stresses obtained between the interface of the implant flange and the crestal, cortical bone. A clear and well-recognised characteristic of the external hexed Brånemark implant is the loss of bone in the period following abutment connection and early loading from the level of the abutment-implant interface down to the first thread of the implant. The aetiology of this is not clear, but a number of hypotheses have been put forward. It has been proposed for example that there is micro-leakage between the abutment and the implant, of the implant-abutment interface, resulting in the release of bacterial toxins causing a local peri-implant reaction and bone loss within a localised zone around this interface.

A third, but not commonly discussed reason may well be related to mechanical stresses. This is particularly prevalent with the earlier designs of implants using commercially pure, type I titanium, which is relatively soft. In such cases the combination of a static load at implant placement around the flange between the implants and the cortical bone and then the superimposed dynamic stresses at the time of implant loading will cause a high level of mechanical stress and bending within the implant body, around the neck, between the flange and the threaded body of the implant. The bone response to these localised stresses is likely to be resorption; the high levels of stress at the implant neck can be visualised by a stress profile, superimposed on an implant in bone in this region.

It is therefore highly desirable to have a flange and neck design that minimises both the static stresses at placement and the dynamic stresses on the functional loading. The Neooss implant has been designed specifically without a neck region, as in the external hex implant designs, and the thread leads directly into the flange. The relationship between the flange and the threaded portion of the implant is slightly different for the 3.5, 4 and 4.5mm diameters. This enables the use of one common abutment connection without impairing the fit of seating of the implant in the surrounding bone.

In the flange region, it is therefore possible to assess the level of compression occurring in the flange during implant insertion. The Neo system has great for a two-stage implant with a high level of surrounding bone a counter sink can be used to provide the optimal seating, minimising static stresses and optimising the interface between the flange and the surrounding bone, thereby providing optimal conditions for direct bone formation. The Neo implant system is therefore designed with a number of features in geometry, preparation technique and material properties that jointly result in the optimal biomechanical relationship between a dental implant and the surrounding bone.

REFERENCES


Ansell R, Scales J (1968). A study of some factors which affect the


Histological Evaluation of a Bimodal Titanium Implant Surface: A Pilot Study in the Dog Mandible

Luiz A Salata¹, Paulo EP Faria¹, Marconi G Tavares¹, Neil Meredith²,³ and Lars Sennerby⁴

¹Dept. Oral & Maxillofacial Surgery, Faculty of Dentistry of Ribeirao Preto, The University of Sao Paulo, Brasil
²University of Bristol, Bristol, UK
³Neoss Ltd, Harrogate, UK
⁴Dept Biomaterials, Institute of Clinical Sciences, Sahlgrenska Academy, Göteborg University, Sweden

This histological pilot study revealed a favourable bone tissue response to a novel bimodal titanium implant surface after four months of healing, with no apparent differences from TiO-blasted and oxidized control implants.

INTRODUCTION

New implant systems with different geometries and surface topographies are continually being launched on the market. It is important to evaluate critically each implant surface in both experimental models and in clinical follow-up studies. One prerequisite for a successful clinical outcome with osseointegrated titanium implants is secure bone integration immediately following surgery (Albrektsson et al. 1981). In essence, the surgical trauma initiates a healing process which includes the formation of a blood clot, migration and differentiation of cells, formation of a granulation tissue and, finally, bone formation and remodelling. In the presence of a titanium surface, healing results in formation of direct bone-implant contacts (BICs) and the number and extent of BICs increase with time (Johansson et al. 1987, Sennerby et al. 1993). The peak torque required to achieve implant removal increases with time in parallel with increased BICs (Johansson et al. 1987). The first generation of osseointegrated implants had either a minimally rough surface (machined/turned surface) or a very rough surface produced by titanium plasma spraying (TPS) (Bränemark et al. 1969, Schroeder et al. 1976). On the basis of further clinical and experimental research it is currently believed that moderately rough implant surfaces are preferable (Albrektsson & Wennerberg 2006); such as surfaces produced by blasting, anodic oxidation, acid etching or combinations of these techniques. Experimental research has generally demonstrated a stronger bone tissue response to surface modified implants than to smoother control surfaces, indicating more rapid integration (Albrektsson & Wennerberg 2006).

The aim of the present pilot study was to analyse the bone tissue response to a novel implant surface (Bimodal surface, Neo Implant System™) in comparison with two well-documented and commercially available implant surfaces (TiO-blasted surface and TiUnite™ surface).

MATERIALS AND METHODS

Implants

A total of eight test implants, 9 mm long and 3.5 mm in diameter (Neo Implant System™, Neoss Ltd, Harrogate, UK) (NE implants) were used in the study. These implants had a bimodal surface created by blasting with 100 to 300 μm diameter ZrO₂ spheres and subsequent blasting with irregularly shaped TiO-based particles, 75 to 150 μm wide (Fig. 1a). Eight control implants were also used; four implants with a TiO₂-blasted surface, 9 mm long and 3.5 mm in diameter (MicroThread™, AstraTech AB, Mölndal, Sweden) (AT implants)(Fig. 1b), four implants with an oxidized surface, 10 mm long and 3.75 mm in diameter (Brånemark System™, MKIII, TiUnite, Nobel Biocare AB, Gothenburg, Sweden) (NB implants)(Fig 1c).

Animals and anaesthesia

Four mongrel male dogs weighing between 20 and 25 kg were used in the study. The animals were preanaesthetized with xilazine (Ronpum®, Brazil, 20 mg/Kg I.M.) and ketamine 1g (Dopalen®, Brazil, 0.8 g/Kg I.M.) and anaesthetized with thionembutal 1 g (Tiopental®, Brazil, 20 mg/Kg I.V.). The animals were kept on intravenous infusion of saline during surgery, all of which was carried out under sterile...
conditions. After surgery the animals received intravenously vitamin compound (Potenay®, Brazil); an anti-inflammatory/analgesic (Banamine®, Brazil) and antibiotic (Pentabiótico®, Brazil). The antibiotic was administered in single doses immediately after surgery, and then 48 and 96 hours postoperatively. The study protocol had been approved by the University of Sao Paulo’s Animal Research Ethics committee.

Experimental protocol
The mandibular premolars were extracted five months prior to commencement of the experiment. At the time of implant placement, crestal incisions were made and mucoperiosteal flaps were raised bilaterally. Two implant cavities were prepared on each site, in accordance with the manufacturers guidelines. Two NE implants were placed on one side and one each of AT and NB implants on the contra lateral side (Figs. 2a and b). Cover screws were placed and the flaps were closed and sutured.

After 4 months of healing, the dogs were sacrificed and the implants with their surrounding tissues were harvested and fixed by immersion in buffered formaldehyde.

Histology
The fixed specimens were dehydrated in a graded series of ethanol and embedded in light curing methacrylate (Technovit® 7200 VCL, Kulzer, Friedrichsdorf, Germany). Ground sections approximately 10 μm thick were prepared using a sawing and grinding technique (Exakt Apparatebau®, Norderstedt, Germany) and stained with toluidine blue. One central section was taken from each implant site in the bucco-lingual direction.

The sections were examined under a Leitz microscope equipped with a Microvid system for morphometrical measurements. The degree of bone-implant contact (BIC) was measured from the first bone contact and expressed as a percentage mean total BIC. The bone area (BA) within the implant threads was measured and expressed as a mean total BA.
RESULTS AND DISCUSSION

Healing was uneventful in all dogs. Some marginal bone resorption, especially on the buccal aspect, was seen for all implant types at the marginal portion which may be due to remodelling continuing after tooth extraction (Arajou et al. 2005).

Histology revealed close contact between mature bone and test implants (Figs 3). There were no apparent differences between test and control implants and new bone was observed to fill the threads of all three surfaces (Figs. 4a-c). In some areas, the presence of a thin layer of bone and an osteoblast seam facing the bone marrow indicated bone formation directly at the test implant surface (Fig. 5); as previously described for the control implant surfaces used in the study, i.e. blasted and oxidized surfaces (Ivanoff et al. 2001, Rocci et al. 2003). It has been speculated that such direct bone formation seen at surface modified implants is the result of an adherent blood clot through which mesenchymal cells can migrate and differentiate to form bone directly on the implant surface (Davies 2003).

The morphometric measurements showed no apparent differences with regard to BIC or BA as similar mean values were obtained (Fig. 5). However, few animals were used and statistics could not be applied.

CONCLUSION

The present pilot study revealed bone integration of the novel bimodal implant surface, with no apparent differences from TiO-blasted and oxidized control implants after four months of healing.
Figure 4. Light micrographs showing bone formation towards all three implant surfaces: a/ Bimodal surface, b/ TiO-blasted surface, c/ Oxidized surface

REFERENCES


Figure 5. Light micrograph of a bimodal implant surface (I). Bone (B) is formed from the surface and towards the bone marrow (BM). Os = osteoid, V= vessel

Figure 6. Results from morphometrical measurements of BIC and BA.


INTRODUCTION

Osseointegrated implants are clinically successful if a direct bone-implant contact can be established and maintained (Albrektsson et al. 1981). The bone-implant interface is biomechanically challenged in rotational, axial and lateral directions during healing, the prosthetic phase and clinical function. The ability to withstand loading is decisive for the clinical outcome and factors of importance are (i) type and magnitude of loading, (ii) the quality of the bone-implant integration and (iii) the mechanical properties of the surrounding bone. Implant integration is time dependent and the biomechanical properties of the bone-implant interface improve with time (Johansson et al. 1987, Sennerby et al. 1993, Friberg et al. 1999). Therefore, the use of a two-stage procedure with three to six months of healing usually ensures a mature bone-implant interface and good clinical results. However, the trend today is to use immediate/early loading protocols, which make great demands on the bone-implant interface since the implants will be loaded during initial healing.

The first generation of osseointegrated implants had a relatively smooth (machined, turned) surface (Brånemark et al. 1969). Good long-term clinical outcomes have been reported on all indications when used in good bone qualities and using a two-stage procedure (Albrektsson & Sennerby, 1991). However, in more challenging situations such as low bone densities, bone grafting and immediate loading, increased failure rates have been reported (Friberg et al. 1993, Becktor et al. 2004, Glauser et al. 2001).

Surface modification is one way of improving implant integration and stability, as shown in numerous experimental studies (Albrektsson & Wennerberg 2004). It is believed that the surface irregularities ensure a firm contact with the blood clot allowing primitive cells to migrate to the interface, differentiate to osteoblasts and form bone directly on the surface (Davies 2003). For a smooth surface, shrinkage of the blood clot will create a gap at the interface and cells cannot reach the surface (Miranda-Burgos et al. 2007). Thus, implants with a moderately rough surface integrate more rapidly and with more bone contacts than smooth surfaced implants (Ivanoff et al. 2001, Zechner et al. 2003).

A second means of improving implant integration is by geometric features. Most dental implants are self-tapping and have an apical configuration including cutting edges and bone chambers. Bone ingrowth into such voids is most likely to increase the rotational stability of the implant. Recent research has indicated that grooves at the thread flank may lead to improved healing by guided bone formation as well as to an improved interlock with bone (Hall et al. 2005). The Neoss implant is a self-tapping implant with apical bone chambers and vertical flutes. It has a bimodal surface topography which is produced by blasting with two different sizes of ZrO and Ti-based particles. The influence of the surface topography and geometry on the bone tissue response and stability is presently not known.

The present study was conducted to examine the early tissue responses to the bimodal titanium surface. The aim was also to evaluate the effect of surface topography and geometrical features on rotational stability.
MATERIALS AND METHODS

Implants

A total of 96 implants, 7 mm long and 4.0 mm in diameter were implanted in the study. These were both original and modified Neoss implants (Neoss Ltd, Harrogate, UK) as follows (fig. 1):

- 48 implants with a bimodal surface created by blasting with 100 to 300 μm wide ZrO₂ spheres followed by irregularly shaped Ti-based particles, 75 to 150 μm wide (Fig 2).
  - 36 with vertical flutes (original surface and geometry) (B+)(Fig 3a)
  - 12 without flutes (B-)(Fig 3b)
- 24 implants with turned surfaces
  - 12 with two vertical flutes (original geometry) (T+)
  - 12 without flutes (T-)
- 24 other implants used for another investigation.

Animals, anaesthesia and experimental protocol

A total of 12 female New Zealand white rabbits were used in the study, after the protocol had been approved by the animal ethics committee of the Gothenburg University. General anesthesia was induced by intramuscular injections of fluanisone-fentanyl (0.7 mL, Hypnorm™, Helsingborg, Sweden) and intraperitoneal injection of diazepam (0.25 mg/kg, Apozepam™, Alpharma AB, Stockholm, Sweden). Local anesthesia was induced at both proximal tibial metaphyses and distal femoral condyles by injections of lidocaine (about 2 mL, Xylocaine®, Astra Zeneca AB, Södertälje, Sweden).

The distal femoral condyles and proximal tibial metaphyses were used as experimental sites. The bone was exposed via incisions through skin and fascia. Two implant sites were prepared in each femur and tibia; each animal receiving 8 implants. Implants were selected by a rotational scheme, so that one each of T+, T-, B+ and B-implants were placed in the femora of each animal. Two B+ and two B- implants were placed in the tibiae of each animal. Cover screws were placed and the flaps were closed with resorbable sutures. The animals were allowed to heal for three (n=4) and six (n= 8) weeks after surgery.

Removal torque

After six weeks of healing, the femoral implants in eight animals were subjected to removal torque (RTQ) test. The tests were performed in a specially designed rig using a motor-driven device. A linearly increasing torque was applied until failure of integration occurred; the peak value in Newton-centimeters (Ncm) was recorded. A mean value was calculated.
for each of the four implant types. The percentage difference from that of turned implants without vertical flutes was calculated for each of the other groups. The Wilcoxon Signed Rank test was used for statistics and a difference was considered if $p < 0.05$.

**Histology**
All implants and surrounding bone tissues were retrieved and fixed by immersion in a 4% buffered formaldehyde solution. The specimens were then dehydrated in a graded series of ethanol and embedded in light curing methacrylate (Technovit® 7200 VCL, Kulzer, Friedrichsdorf, Germany). Ground sections, approximately 10 μm thick, were prepared using a sawing and grinding technique (Exakt Apparatebau®, Norderstedt, Germany). One central section was taken from each implant site, stained with Toluidine Blue and examined in a Nikon light microscope.

**RESULTS**
Light microscopy of the three-week specimens of the bimodal implant surface revealed bone formation directly onto the implant surface (Fig. 4a). This could be seen as thin rims of bone following the contour of the implant threads (Figs. 4a and b) and as solitary islets with no obvious connection to existing bone surfaces (Fig. 5). Osteoblastic seams were often seen on the bone rims, facing the adjacent soft tissues (Figs. 4b and 6). Non-bone areas consisted of a loose connective tissue rich in cells and vessels and devoid of signs of inflammation (Figs. 4a-b, 5 and 6). A more mature bone-implant interface was seen in the six-week specimens and a larger proportion of the implant surface was in contact with bone (Fig. 7).

The removal torques were found to be correlated with surface topography and the absence or presence of vertical flutes (Fig. 8); the lowest torques were recorded for machined implants without flutes and the highest for bimodal surface implants with vertical flutes ($p<0.05$)(TABLE 1).

**DISCUSSION**
The present study indicated that the bimodal topography induced bone formation directly at the implant surface as previously described for other commercially available surface modified implants (Piattelli et al. 1996, Ivanoff et al. 2001, Rocci et al. 2003, Berglundh...
et al. 2003). This was seen as seen thin rims or solitary islets of bone at the surface with osteoblastic seams facing the adjacent bone marrow. Previous descriptions of the healing of smooth, turned implants have reported bone formation from the adjacent tissues and towards the implant surface (Sennerby et al. 1992, Palma et al. 2006, Miranda-Burgos et al. 2007). The mechanisms behind the different integration pathways are probably related to the integrity of the blood clot-implant interface during the early events of bone healing (Davies 2003). With a rough surface, the clot can maintain a firm contact in spite of shrinkage, whereas a gap may be formed at a smooth implant surface. In the former case mesenchymal cells can migrate to the implant surface, differentiate and start to produce bone matrix. The influence of surface modification on the clinical outcome is not clear. For instance, clinical studies comparing titanium-plasma sprayed and turned surfaces or TiO2 blasted and turned surfaces could not find any statistically significant differences with regard to survival rate and marginal bone loss (Astrand et al. 2004a, 2004 b). However, other non-comparative studies have indicated better survival rates for surface modified than for turned implants in challenging clinical situations such as bone grafting (Brechter et al. 2005) and in immediate loading (Glauser et al. 2001, 2003). Since the trend today is to use immediate/early loading, the use of surface modified implants is preferable.

The removal torque tests revealed an improved resistance to torque with modified topography and a vertical flute added. This is best explained by ingrowth of bone into micro- and macroscopic undercuts at the surface. Hall et al (2005) showed that a macroscopic groove added to the thread flank can stimulate bone formation over the implant surface. The relatively

<table>
<thead>
<tr>
<th>Turned</th>
<th>Turned with flute</th>
<th>Bimodal</th>
<th>Bimodal with flute</th>
</tr>
</thead>
<tbody>
<tr>
<td>(T-)</td>
<td>(T+)</td>
<td>(B-)</td>
<td>(B+)</td>
</tr>
<tr>
<td>42.4</td>
<td>44.4</td>
<td>46.9</td>
<td>58.5</td>
</tr>
<tr>
<td>(15.4)</td>
<td>(15.0)</td>
<td>(13.2)</td>
<td>(13.2)*</td>
</tr>
</tbody>
</table>

* P<0.05 compared with T- implants

Table 1. Results from removal torque measurements.

Figure 5. Light micrograph after three weeks showing solitary bone formation (B) at the apical part of the implant (I) facing a loose connective tissue (LCT) rich of cells.

Figure 6. Light micrograph showing bone formation at the bottom of a thread after three weeks of healing. New bone (NB) is formed by osteoblasts (Os) on previously formed bone (PB) and separated by a cement line (white arrow). LCT = loose connective tissue.

Figure 7. Light micrographs after 6 weeks of healing showing almost complete bone filling of a thread with mature bone. Arrows point at newly formed secondary osteons indicative of remodelling.
small effect of surface topography may be explained by the design of the test implants, since they all had apical undercuts, i.e. cutting edges and bone chambers.

CONCLUSION

The present experimental study showed evidence of surface mediated bone formation at the bimodal surface as previously described for other commercially available surface modified implants. Removal torque tests showed increased stability with the modified surface and adding vertical flutes when compared to turned control implants without flutes.

REFERENCES


Miranda-Burgos P, Rasmusson L, Meirelles L, Sennerby L. Early bone tissue responses to oxidized and machined titanium implants in the rabbit tibia. Clin Implant Dent Relat Res. Accepted for publication


A One-Year Clinical, Radiographic and RFA Study of Neoss Implants Used in Two-Stage Procedures

Peter Andersson¹, Damiano Verrocchi¹, Rauno Viinamäki¹, Lars Sennerby¹,²

¹Private Practice, Fiera di Primiero and Feltre, Italy
²Dept Biomaterials, Inst Clinical Sciences, Sahlgrenska Academy, Gothenburg University, Sweden

This study reports a survival rate of 98.1% for 102 Neoss implants in 44 patients with a mean bone loss of 0.7 mm after one year. The reduced component inventory and innovative designs enhancing primary stability and facilitating laboratory technology/prosthetics offer practical advantages whilst the clinical results with the implant system tested compare favourably with existing systems.

INTRODUCTION

The use of implant-supported bridges is a routine treatment modality for the edentulous patient with documented good long-term results (Albrektsson & Sennerby 1991, Esposito et al. 1998). From initially a rather complicated and restricted procedure, the techniques of implant-supported dentistry have improved and simplified, at least from a surgical point of view. The introduction of self-tapping and surface modified implants, surgical guides, shortened healing periods and immediate loading are some examples. However, improvements could still be made in dental laboratory and prosthetic techniques in order to reduce further treatment times and the overall cost of the treatment.

The aim of the present study was to evaluate a new implant system (Neoss) for one year using clinical and radiographic examinations and implant stability measurements by resonance frequency analysis (RFA).

MATERIALS AND METHODS

Patient inclusion

Consecutive patients requiring implant treatment for total or partial loss of teeth were included in the study until at least 100 implants had been inserted. Only two-stage procedures with 3 months of healing from placement to abutment connection were performed.

Preoperative examinations included intraoral and panoramic radiographs. Computerized tomography was used if required. The implant sites should have sufficient bone for at least 7 mm long implants. Patients should be over 18 years old and present no contraindications for oral surgery under local anaesthesia.

Clinical techniques

Patients were administered 2 gr of amoxicillin (Augmentin™, Roche, Milan, Italy) and sedation if required (Valium®, Roche, Milan, Italy, 5 mg) prior to surgery. Anaesthesia was induced by infiltration with articain/epinephrine (Septocain™, Specialites Septodont, Saint-Maur-Des-Fosses, France). Crestal incisions were used for flap elevation. Implant sites were prepared with a 2.2 mm twist drill followed by a 3 mm (3.5 mm wide implant), 3.4 mm (4.0 mm wide implant) and 3.9 mm drills (4.5 mm wide implant) (Neoss ltd, Harrogate, UK). Full countersink preparation was made for all implants. Implants were inserted with the torque driver set to 40 Ncm and final seating was performed with a hand wrench. Cover screws were applied and the flaps were replaced and sutured. Bone quality and quantity according to Lekholm and Zarb (1985) were registered.

Abutment connection was performed 3 months later, usually with a punch technique and no sutures. Bite registration and impressions were taken with closed tray and then healing abutments were connected. Neolink™ abutments (Neoss ltd, Harrogate, UK) of gold or titanium (to match the metal of the framework) were used. Abutments were either cast or welded into the metal frameworks. Porcelain and acrylic veneers were used. Constructions were screw-retained if implant angulation permitted; otherwise, individual abutments (Neo Matrix, Neoss ltd, Harrogate, UK) were used for cementation.
Follow-up
Implant stability was assessed by RFA (Mentor®, Integration Diagnostics AB, Gothenburg, Sweden) at implant surgery, abutment connection and after one year of loading (after unscrewing the constructions). Cemented constructions were not removed.

Digital or conventional intraoral radiographs were taken at abutment connection and after one-year of service. Conventional radiographs were photographed with a digital camera on a light desk. Measurements were made using a personal computer (Image J 1.34S, National Institutes of Health, USA) at mesial and distal aspects. Each radiograph was calibrated using the known width of the coronal cylinders of the implants. The upper platform was used as a reference point for measurements (Figure 1).

RESULTS
Forty-four (44) patients (16 male and 28 female, mean age 54 years) were treated according to the protocol. Forty-eight (48) prosthetic constructions were delivered including five full cross-arch bridges, 28 partial bridges and 17 single crowns. A total of 102 implants were placed, 34 in the maxilla and 68 in the mandible, in bone quality and quantity as presented in Table 1. Implant lengths and diameters are presented in Tables 2 and 3.

At writing all patients have completed one year in function. All constructions except one patient with a cemented bridge on five implants were removed for stability checking.

All patients maintained a fixed bridge or crown during the one year study. Two failures were experienced, giving a survival rate of 98.1% after one year. One 13 mm long by 4.0 mm wide implant placed in a defect in a mandible was removed at impression due to mobility. In this case, a new implant was placed and a temporary bridge was made on the remaining three implants whilst awaiting healing. One 7 mm long by 4.5 mm wide implant in Q4 bone in the 1st molar region in a maxilla was lost after one year. The implant was removed and the three-unit bridge was supported on the remaining two implants. No other prosthetic complications such as fractures or screw-loosening were experienced.

A total of 92 implants with readable radiographs at
baseline and follow-up have been evaluated to date. On average, the marginal bone level was located 0.6 mm (SD 0.7) below the platform at baseline and 1.3 mm (SD 1.0) below after one year; giving an average bone loss of 0.7 mm (SD 1.0) at one year. X implants showed more than 3 mm bone loss.

Stability measures by RFA gave a mean of 75.0 (SD 6.5) ISQ at placement, 74.4 (SD 6.8) at abutment connection and 76.2 (SD 6.6) after one-year. There seems to be a relation between bone quality and ISQ at implant placement (Fig. 4). Implant stability increased with time for implants in Q4 bone whilst stability decreased for implants in very dense bone, resulting in a similar stability for all implants after one year.

DISCUSSION

Early experience with the Neoss implant system used in a two-stage procedure with 3 months of healing in the present study shows satisfactory results. Two of 102 implants were lost during one year which compares well with the results reported from other implant systems (Albrektsson & Wennerberg 2004). From a surgical point of view, the implant was stable during insertion and good primary stability was generally achieved. When poor primary stability was encountered, a wider implant was used to enhance stability. Since the different implant diameters have the same prosthetic platform, an implant could be replaced with a wider diameter without requiring changes to the other prosthetic components, as is commonly necessary with other implant systems. Abutment connection was easier with this implant than previously experienced with external connection implants, which often require a flap procedure and removal of bone tissue in order to fit an abutment. A punch technique without suturing could frequently be used in the present study and in most cases impressions were taken in conjunction with abutment connection. The use of integrated abutments with the framework facilitated prosthetics and reduced the overall costs.

The RFA measurements revealed a high average primary stability of these implants which is most probably due to the geometric features of the implant. Firstly, it has a two-start thread, a twin helix, with two threads running parallel but on diametrically opposite sides of the cylindrical substrate. Thus, the actual thread pitch is double the “apparent” thread spacing, so the implant advances twice as far with one turn as for a single-helix thread. This feature reduces the insertion time and counteracts wobbling during insertion. Secondly, the implant has a positive tolerance, meaning that the implant is slightly conical in the coronal direction and as the implant screws home its diameter increases slightly and there is a slight tendency to tighten the thread base against the bone. Previous research has shown higher stability for slightly tapered implants in soft bone than for parallel-sided implants, without jeopardizing the integration process (O’Sullivan et al 2000, O’Sullivan et al 2004a, 2004b). Decreased primary stability was seen with decreased bone density, which is in line with the findings of Östman et al (2006). Implant stability increased slightly with time and was similar for all bone qualities after one year of function, which indicates a favourable bone tissue response to the implants. The greatest increase of stability was seen for implants in Q4 bone which is in line with the findings of Friberg et al. (1999). This can be explained by stiffening of the bone-implant interface with time due to bone formation and remodelling.
In contrast, decreased stability was observed in dense bone, which may be explained by mechanical relaxation and/or bone remodelling as a response to the presumably high stresses induced by implant placement.

The radiographic analysis revealed that about 0.7 mm of bone was lost during the first year of loading, which is in the range of previously reported results with other systems (Oh et al. 2002). X implants showed more than 3 mm bone loss during one year. The fact that the bone level was located 0.6 mm below the platform at abutment connection indicates some remodelling during healing as generally the implants were placed flush with the surrounding bone. This accords with the findings of Engquist et al. (2001) who compared the marginal bone tissue response at Astra Tech and Brånemark system implants.

It is concluded that prosthetic rehabilitation of the edentulous patient with the Neoss implant system results in good short-term clinical and radiographic outcomes. The innovative design solutions do seem to enhance primary stability and facilitate laboratory technology/prosthetics, and with the reduced component inventory this system offers advantages over our experiences with other implant systems.

REFERENCES


Immediate/Early loading of Neoss Implants. Preliminary Results from an Ongoing Study

Peter Andersson¹, Damiano Verrocchi¹, Luca Pagliani², Lars Sennerby³

¹Private Practice, Fiera di Primiero and Feltre, Italy
²Private Practice, Milano and Legnano, Italy
³Dept Biomaterials, Inst Clinical Sciences, Sahlgrenska Academy, Gothenburg University, Sweden

This interim report of an ongoing study on immediate/early loading of Neoss implants reports a survival rate of 96.5% for 141 Neoss implants in 33 patients after 6 months to three years of loading. All patients received and maintained a fixed bridge in spite of five failures in three patients.

INTRODUCTION

The use of immediate/early loading protocols for implant-supported crowns and bridges has obvious advantages for the patients since only one surgical procedure and no healing periods are needed. Both function and aesthetics can be immediately restored with a temporary crown or bridge. Concerns have been raised about increased failure rates, since the original concept of osseointegration prescribed a submerged and unloaded healing period of 3 to 6 months before loading (Brånemark et al. 1969, Albrektsson et al. 1981). Today, histology from experimental and clinical studies has demonstrated that implants can integrate under the influence of functional loads (Piattelli et al. 1998, Rocci et al. 2003). Moreover, clinical follow-up studies have reported similar good clinical outcomes as for two-stage procedures (Attard & Zarb 2003). In fact, immediate/early loading is now a clinical reality and a commonly used procedure. In many studies, authors have identified primary implant stability as a critical factor for success (Östman et al 2006). Inclusion criteria based on insertion torque values and RFA measurements have been used (Calandriello et al, 2003, Rocci et al 2003, Vanden Bogaerde et al 2004, Östman et al 2005). It has been shown that the use of self-tapping implants, reduced final drill diameters and tapered implants may improve primary stability (O’Sullivan et al 2001, 2004).

The aim of the present study was to report on the early experiences of an immediate/early loading protocol using a new implant designed to give firm primary stability.

MATERIALS AND METHODS

Patient group

The study group consisted of 33 patients (18 female and 15 male) from three clinics, representing consecutive treatments with immediately/early loaded implant-supported provisional bridges or crowns. The patients were totally edentulous (n=18), partially edentulous (n=12) or treated for single tooth loss (n=3) (Table 1). The possibility of placing long implants with good primary stability was assessed in each patient on the basis of radiography and clinical examinations. A final decision was taken after discussions with the patient.

Implant surgery was performed under local anaesthesia and a total of 141 implants were inserted, 79 in the maxilla and 62 in the mandible (Neoss Ltd, Quantity Number of sites Quality Number of sites

<table>
<thead>
<tr>
<th>Quantity</th>
<th>Number of sites</th>
<th>Quality</th>
<th>Number of sites</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>10</td>
<td>1</td>
<td>3</td>
</tr>
<tr>
<td>B</td>
<td>104</td>
<td>2</td>
<td>81</td>
</tr>
<tr>
<td>C</td>
<td>26</td>
<td>3</td>
<td>48</td>
</tr>
<tr>
<td>D</td>
<td>1</td>
<td>4</td>
<td>9</td>
</tr>
<tr>
<td>E</td>
<td>-</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 1. Bone quantity and quality of implant sites according to Lekholm and Zarb (1985).
Harrogate, UK) using an insertion torque of at least 40 Ncm. Primary stability was ensured by reducing the final drill diameter in soft bone qualities. The majority of implants were 4 mm in diameter and 13-15 mm long (Table 2 and 3). Most implants had been placed in bone quantity b and quality 2 according the Lekholm & Zarb index (Table 4 and 5).

Sterile impression copings were attached directly to the implants (n=75) or to straight or angulated abutments (n=66) (Southern Implants™, Protera AB, Gothenburg, Sweden). Impressions and bite registration were taken after suturing. Healing abutments were then connected to the implants. A screw-retained laboratory-made crown (n=3) or metal-reinforced bridge (n=31) was delivered one to three days after surgery. In two patients treated in the posterior mandible, a provisional bridge was made in the mouth the same day. Altogether, 34 provisional prosthetic constructions were delivered to the 33 patients. Single and partial constructions were not in occlusion. Total cross-arch bridges were carefully adjusted with regard to occlusion, striving towards group function and contacts over implant sites.

After a period of 3 to 6 months, the provisional constructions were replaced with permanent crowns or bridges made of gold or titanium with acrylic or porcelain veneers.

**Follow-up**

The patients were monitored with clinical examinations controlling the occlusion during the first weeks and months. The protocol includes RFA measurements (Mentor®, Integration Diagnostics AB, Gothenburg, Sweden) at implant surgery, on delivery of the final prosthesis and after one year of loading.

Digital or conventional intraoral radiographs were taken at baseline and after one-year of service.

### RESULTS AND DISCUSSION

All 33 patients have passed 6 months of loading and 13 have exceeded one year. A total of five failures were experienced, giving a survival rate of 96.5% after 6
months to 3 years. All patients received and maintained a fixed prosthesis during the follow-up period. The failures all occurred in the maxilla within three months of surgery (Table 4). Implant stability was 73.5 (SD 7.0) ISQ at placement and 74.4 (SD 9.2) ISQ at final prosthesis delivery after 4 months on average.

DISCUSSION

The use of immediate/early loading protocols conveys obvious advantages for the patient since only one surgery and no healing periods are needed. Moreover, there is no need for removable appliances which can be difficult and uncomfortable to wear. In the present study, a fixed provisional bridge or crown was fabricated chair-side or delivered within a few days. In this interim report of an ongoing study, five of 141 implants failed after a follow-up of at least 6 months. All patients received and maintained a fixed construction throughout the study period in spite of five failed implants. The 3.5 % failure rate of this study is similar to those reported by other authors in studies on immediate loading. For instance, Glauser et al lost about 3 % of implants when used on all indications as in the present study. All failures in the present study occurred in the maxilla; a failure rate of 6.3 % which corroborates the findings of Olsson et al (2003). Consequently, the results were better in mandibular bone as no implant failures were experienced; which is also in line with the findings in the literature (Attard & Zarb 2005). It can be speculated that differences in bone density may explain this finding as implants get better primary stability in mandibular bone (Östman et al 2006). However, all but one implant showed high stability values and other factors such as unfavourable loading need to be considered. The implants used in the present study have a surface topography modified by blasting. It is known from animal experiments that such surfaces seem to integrate faster and with more bone contacts than implants with a smooth surface. It is also possible that this contributed to the good outcome of the present study. The RFA measurements revealed firm primary stability

This interim report of an ongoing study on immediate/early loading of Neoss implants reports a survival rate of 96.5 % for 141 Neoss implants in 33 patients after 6 months to three years of loading. All patients received and maintained a fixed bridge in spite of five failures in three patients.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Position</th>
<th>Case</th>
<th>Diameter/Length</th>
<th>Quantity/Quality</th>
<th>Primary stability (ISQ)</th>
<th>Time of failure</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>16</td>
<td>Partial</td>
<td>4/15</td>
<td>C/3</td>
<td>44</td>
<td>3 months</td>
</tr>
<tr>
<td>2</td>
<td>13</td>
<td>Total</td>
<td>4/13</td>
<td>B/3</td>
<td>75</td>
<td>3 months</td>
</tr>
<tr>
<td>3</td>
<td>14</td>
<td>Total</td>
<td>4/13</td>
<td>B/2</td>
<td>75</td>
<td>3 months</td>
</tr>
<tr>
<td>3</td>
<td>11</td>
<td>Total</td>
<td>4/13</td>
<td>B/2</td>
<td>66</td>
<td>3 months</td>
</tr>
<tr>
<td>3</td>
<td>26</td>
<td>Total</td>
<td>4/15</td>
<td>B/3</td>
<td>77</td>
<td>3 months</td>
</tr>
</tbody>
</table>

Table 4. Characteristics of implant failures.
REFERENCES


A Retrospective Follow-up of 50 consecutive patients treated with Neoss Implants with or without an Adjunctive GBR-Procedure

INTRODUCTION

The use of osseointegrated implants has been proven to result in good clinical outcomes in the prosthetic rehabilitation of the edentulous patient, whether used in totally or partially edentulous jaws, including single tooth replacements (Esposito et al. 1998). Since the introduction some 40 years ago, the osseointegration technique has been continuously developed and refined in order to simplify and shorten implant treatment. For instance, the introduction of self-tapping implants, surface modifications and immediate/early loading protocols has markedly facilitated implant treatment. The presence of sufficient bone volumes was originally an absolute criterion for using implants. Today, numbers of reconstructive techniques including the use of bone grafts, bone substitutes, osteodistraction devices and membranes are available to increase the load-bearing volume of the jaw bone. A localized defect such as that due to incomplete healing after extraction may complicate the placement of an implant by exposure of parts of the implant. In such cases, the use of a bovine bone substitute and a resorbable membrane is commonly used to achieve complete bone coverage (Hurzeler et al. 1998, Hammerle & Lang 2001).

The Neoss implant system (Neoss Ltd, Harrogate, UK) is a novel design which according to the manufacturer was developed to provide simple and effective solutions for all kinds of cases with minimal components. Special attention has been paid to the technical/prosthetic phase and only one prosthetic platform is used irrespective of implant diameter.

The aim of this retrospective study was to report on the experiences with Neoss implants from the first consecutive 50 patients treated in one private office.

MATERIALS AND METHODS

Patients and surgery
Fifty consecutive patients (20 males, 30 females, mean age 57 years) needing implant treatment were enrolled in the study. Three patients were treated in both maxilla and mandible resulting in 53 treatment areas (jaws) included in the study. Intraoral and panoramic radiographs as well as CT scans were used for presurgical evaluations. Nine patients were totally edentulous (two in both jaws), 21 were partially edentulous (one in two areas) and 20 patients were treated for single tooth loss.

The patients were administered antibiotics prior to surgery (Dalacin® C 300mg, Pfizer). Surgery was performed under sterile conditions and local anaesthesia with Ultracain® DS Forte. Crestal incisions were used and implant sites were drilled in accordance with the guidelines given by the manufacturer for the appropriate implant diameter. Implants were inserted into position with a drilling unit. A total of 183 Neoss implants (Neoss Ltd, Harrogate, UK) were placed; 116 in the maxilla and 67 in the mandible (Table 1). Implant lengths and diameters are shown in Table 2 and 3. Bone quality and quantity according to the Lekholm and Zarb index (1985) were registered (Tables 4 and 5). The implant collar was either fully submerged in bone (n=1087) or to half its length (n=75).

Thirty of the treatment areas with 126 implants underwent GBR using BioOss™ and a resorbable BioGide™ membrane (Geistlich, Switzerland) simultaneously with implant placement.

Healing abutments were connected after a healing period of 3 to 6 months in 32 treatment areas (89 implants). In 21 areas (94 implants) healing abut-
ments were placed in conjunction with implant surgery and 8 of these (57 implants) were loaded with a crown/bridge for immediate function.

**Prosthetics**

Impressions were taken on implant level for screw-retained prosthetics using Neolink abutments (Neoss Ltd, Harrogate, UK) and gold-ceramic or gold-acrylic frameworks. All crowns and partial bridges were gold-ceramic reconstructions and all but one of the full jaw bridges were gold-acrylic reconstructions. The crowns/bridges were attached with gold screws using a preload of 32 Ncm.

**Follow-up**

The patients were scheduled for annual check-ups with clinical and radiographic examinations using intraoral or panoramic radiographs. Marginal bone level measurements were performed by an independent radiologist in one baseline and one follow-up radiograph.

**RESULTS**

Forty-nine of the patients attended the first annual check up and 26 the second. A total of eight implant failures were registered, all during the first year in service, giving an overall Cumulative Success Rate (CSR) of 95.6% (Table 6). All but one failure occurred in the GBR group, giving a CSR of 94.4% for the GBR-group and 98.2% for the non-GBR group after 1 year (Table 6). The characteristics of failed implants are shown in Table 7. Failures occurred

<table>
<thead>
<tr>
<th>MAXILLA</th>
<th>28</th>
<th>27</th>
<th>26</th>
<th>25</th>
<th>24</th>
<th>23</th>
<th>22</th>
<th>21</th>
<th>11</th>
<th>12</th>
<th>13</th>
<th>14</th>
<th>15</th>
<th>16</th>
<th>17</th>
<th>18</th>
<th>TOTAL</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0</td>
<td>1</td>
<td>11</td>
<td>16</td>
<td>17</td>
<td>9</td>
<td>6</td>
<td>4</td>
<td>4</td>
<td>7</td>
<td>9</td>
<td>12</td>
<td>12</td>
<td>7</td>
<td>1</td>
<td>0</td>
<td>116</td>
</tr>
<tr>
<td></td>
<td>0</td>
<td>3</td>
<td>9</td>
<td>5</td>
<td>3</td>
<td>4</td>
<td>5</td>
<td>1</td>
<td>2</td>
<td>6</td>
<td>5</td>
<td>5</td>
<td>6</td>
<td>11</td>
<td>2</td>
<td>0</td>
<td>67</td>
</tr>
<tr>
<td>MANDIBLE</td>
<td>38</td>
<td>37</td>
<td>36</td>
<td>35</td>
<td>34</td>
<td>33</td>
<td>32</td>
<td>31</td>
<td>41</td>
<td>42</td>
<td>43</td>
<td>44</td>
<td>45</td>
<td>46</td>
<td>47</td>
<td>48</td>
<td>67</td>
</tr>
</tbody>
</table>

Table 1. Distribution of implants in relation to position

<table>
<thead>
<tr>
<th>IMPLANT LENGTH</th>
<th>PLACED IMPLANTS</th>
<th>FAILED IMPLANTS IN TOTAL</th>
<th>FAILED IMPLANTS IN GBR PATIENTS</th>
<th>FAILED IMPLANTS IN NON-GBR PATIENTS</th>
</tr>
</thead>
<tbody>
<tr>
<td>7 mm</td>
<td>7</td>
<td>2</td>
<td>2</td>
<td>0</td>
</tr>
<tr>
<td>9 mm</td>
<td>51</td>
<td>4</td>
<td>3</td>
<td>1</td>
</tr>
<tr>
<td>11 mm</td>
<td>75</td>
<td>2</td>
<td>2</td>
<td>0</td>
</tr>
<tr>
<td>13 mm</td>
<td>44</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>15 mm</td>
<td>6</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>17 mm</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Total</td>
<td>183</td>
<td>8</td>
<td>7</td>
<td>1</td>
</tr>
</tbody>
</table>

Table 2. Placed and failed implants in relation to length

<table>
<thead>
<tr>
<th>IMPLANT DIAMETER</th>
<th>PLACED IMPLANTS</th>
<th>FAILED IMPLANTS IN TOTAL</th>
<th>FAILED IMPLANTS IN GBR PATIENTS</th>
<th>FAILED IMPLANTS IN NON-GBR PATIENTS</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.5 mm</td>
<td>58</td>
<td>5</td>
<td>4</td>
<td>1</td>
</tr>
<tr>
<td>4.0 mm</td>
<td>112</td>
<td>2</td>
<td>2</td>
<td>0</td>
</tr>
<tr>
<td>4.5 mm</td>
<td>13</td>
<td>1</td>
<td>1</td>
<td>0</td>
</tr>
<tr>
<td>5.5 mm</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Total</td>
<td>183</td>
<td>8</td>
<td>7</td>
<td>1</td>
</tr>
</tbody>
</table>

Table 3. Placed and failed implants in relation to diameter

<table>
<thead>
<tr>
<th>BONE QUALITY</th>
<th>PLACED IMPLANTS</th>
<th>FAILED IMPLANTS IN TOTAL</th>
<th>FAILED IMPLANTS IN GBR PATIENTS</th>
<th>FAILED IMPLANTS IN NON-GBR PATIENTS</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>44</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>B</td>
<td>103</td>
<td>4</td>
<td>4</td>
<td>0</td>
</tr>
<tr>
<td>C</td>
<td>26</td>
<td>4</td>
<td>3</td>
<td>1</td>
</tr>
<tr>
<td>D</td>
<td>10</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>E</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>Total</td>
<td>183</td>
<td>8</td>
<td>7</td>
<td>1</td>
</tr>
</tbody>
</table>

Table 5. Placed and failed implants in relation to bone quantity
more often for short implants and in soft bone. In spite of the failures, all patients received and maintained a fixed crown/bridge during the follow-up.

A total of 270 radiographs could be used for measurements of marginal bone levels. On average, the bone was situated 1.6 (S.D. 0.75) mm below the top of the collar at baseline and 1.9 (S.D. 0.73) mm at the follow-up one year later. Thus, the total bone loss amounted to 0.3 (S.D. 0.88) mm (Table 8) during this year. There were only small differences between the GBR and non-GBR groups.

**DISCUSSION**

The experiences with the Neoss implant system from the first 50 patients treated in one clinic are reported. A survival rate of 98.2% was achieved after one to two years in routine cases with no need for bone augmentation procedures. This is in accordance with recent studies on other implant systems (Albrektsson & Wennerberg, 2004). The implant that failed in this group was of 3.5 mm diameter with length 9 mm placed in quality type 4 bone in the maxilla. It was one of 8 implants placed with an immediate loading protocol for a full bridge reconstruction. A higher failure rate was seen when implants were placed with a simultaneous GBR procedure using bovine bone and a resorbable membrane, as 7 of 126 implants were lost (5.6%). The results indicate that osseointegration of the exposed parts of the implants was not achieved for these implants during healing prior to loading, resulting in an unfavourable biomechanical situation. This notion is further supported by the fact that short implants (7 and 9 mm) and implants in soft bone quality failed more often. It is possible that a prolonged healing period is needed, as also suggested for sinus lift procedures with bovine bone (Hallman et al. 2005).
Due to the retrospective character of the present study, radiographs could not be provided for all implants. Prospective studies with radiographs of consecutive implants are needed to properly evaluate marginal bone levels. Nevertheless, the marginal bone measurements indicated an acceptable degree of bone loss during follow-up. The average marginal bone level at follow-up was still situated at the implant collar. For the Brånemark implant design, the marginal bone level usually ends up at the first thread some 1.5 to 2 mm below the platform (Oh et al, 2002). Other implant designs have shown only minor bone loss which may be due to micro threads on the collar (Shin et al. 2006). Too few radiographs were available to evaluate the influence of countersink depth of the implant collar on marginal bone loss.

The clinical experiences with the present implant system from a surgical, prosthetic and laboratory technician point of view were positive. The implant design resulted in firm primary stability in all bone qualities. This is probably due to the geometry of the implant, which has a positive tolerance and is thereby slightly tapered. During insertion the bone is compressed in a lateral direction which increases the stability of the implant. Internal connection is an advantage as it makes abutment connection easy. Impressions are taken on implant level and the technician chooses the type of abutment, which in case of screw-retained prosthetics is integrated with the framework.

### CONCLUSIONS

Within the limitations of the present retrospective study, it is concluded that the Neoss implant system results in good clinical outcomes in routine cases as evidenced by survival rate and marginal bone loss. GBR procedures involving short implants and soft bone seem to increase the risk of implant failure.

![Figure 1. Periapical radiographs taken after delivery of a two-unit bridge and after one year of loading.](image)

**REFERENCES**


Hurzeler MB, Kohal RJ, Naghshbandi J, Mota LE, Conradt J,Hur- 
macher D, Caffesse RG. Evaluation of a new biodegradable barrier to 
facilitate guided bone regeneration around exposed implant threads. 

Lekholm U, Zarb GA. Patient selection. In: Brånemark P-I, Zarb GA, 
Albrektsson T, eds. Tissue integrated prostheses. Osseointegration in 

Oh TJ, Yoon J, Misch CE, Wang HL. The causes of early implant bone 

Shin YK, Han CH, Heo SJ, Kim S, Chun HJ. Ra-
diographic evaluation of marginal bone level around 
implants with different neck designs after 1 year.
**Insertion Torque Measurements During Placement of Neoss Implants**

Luca Pagliani1, Lars Sennerby2, Peter Andersson3, Damiano Verrocchi3, Neil Meredith4

1Private Practice, Milano and Legnano, Italy
2Dept Biomaterials, Inst Clinical Sciences, Sahlgrenska Academy, Gothenburg University, Sweden
3Private Practice, Fiera di Primiero and Feltre, Italy
4Bristol Dental Hospital and University, Bristol, UK

*This clinical study demonstrates that the Neoss implant design develops continuously increasing insertion torque (IT) during placement; as expected for a tapered implant design. This indicates that the total implant length laterally compresses the adjacent bone, providing firm primary stability of the implant. Moreover, the study demonstrates a correlation between insertion torque and resonance frequency analysis measurements.*

**INTRODUCTION**

Primary stability is considered a key factor for the clinical success of dental implants. It is determined by the density of the bone at the site, the surgical technique and the design of the implant (Sennerby & Meredith 1998). Historically, increased failure rates have been reported in sites of low bone density (Jaffin & Berman 1991, Friberg et al 1991). Modified drilling protocols have been proposed with the final drill diameter reduced in an attempt to increase compression and thereby the stability of the implant during insertion (Friberg et al 1999).

Tapered implant designs have been introduced on the market in order to improve primary stability. Experimental and clinical studies with insertion torque (IT) measurements and/or resonance frequency analysis (RFA) have demonstrated higher primary stability for tapered implants compared with parallel-walled implants (O’Sullivan et al 2000, O’Sullivan et al 2004a, 2004b). Parallel-sided implants may show a high peak insertion torque which indicates a high degree of stability. However, this is mainly due to the clamping effect when the implant head reaches the marginal bone, whilst the threaded part of the implant shows little resistance during insertion. Tapered implants demonstrate continuously increasing insertion torque due to lateral compression of the bone from the whole implant length during insertion (O’Sullivan et al 2000). Then stresses would be distributed along the tapered implant surface and not concentrated to a few spots.

According to the manufacturer, the Neoss implant has a positive tolerance, meaning that it is slightly conical in the coronal direction, like a tapered implant. However, the insertion torque characteristics of that implant design are not known at present. The objective of this clinical study was to evaluate the primary stability of the Neoss implant using IT and RFA measurements. The aim was also to look for correlations between IT and factors such as bone quality and RFA.

**MATERIALS AND METHODS**

A total of 118 implants (Neoss ltd, Harrogate, UK) of different lengths and diameters (Table 1) placed in both jaws (59 mandibular, 59 maxillary) of 38 patients were evaluated at placement surgery. Insertion torque in Ncm was measured at 20 rpm and 8 Hz to a maximum of 50 Ncm using a drilling unit specially designed for implant surgery (Elcomed, W&H, Milano, Italy)(Figure 1). A torque/time curve was obtained, saved and extracted. The final drill diameter and the degree of countersinking were noted. After final seating, the stability of each implant was measured with resonance frequency analysis in ISQ units (Mentor, Ossell AB, Gothenburg, Sweden). Bone density and quantity were assessed by the Lekholm & Zarb index.

The torque/time curves were examined for mean insertion torque over the total curve and for the apical (E1), mid (E2) and coronal thirds (E3). The E1 and E3 parts always included 40 measurements (corresponding to 5 seconds). The registration time for the E2 part varied due to the different implant lengths accounted for.
Spearman’s rho test was used to test for possible correlations. A statistically significant correlation was considered if \( p<0.05 \).

RESULTS

The torque/time curves displayed continuously increasing torque during insertion (Figure 2). Thus, the implants behaved as expected for a tapered implant design as previously described by O’Sullivan et al (2000). The total mean insertion torque was 15.1 Ncm (SD 7.2) and for regions was 4.8 Ncm (SD 6.5) in the E1, 13.3 Ncm (SD 6.5) during E2 and 26.9 Ncm (SD 11.9) in E3 (Figure 3). Except for the E1 region, these torques are higher than those previously reported by Friberg et al (1999a) in 523 self-tapping parallel-sided implants; used in both jaws of 105 patients. The data indicate higher primary stability for the present implant design as described by IT measurements. This accords with previous experimental research which showed higher stability for tapered implants than for parallel-sided ones, without jeopardizing the integration process (O’Sullivan et al 2000, O’Sullivan et al 2004a, 2004b).

A statistically significant correlation between IT and RFA was demonstrated for the total implant lengths as well as for the E1-E3 regions individually (Table 2). The same correlation was reported by Friberg et al (1999b) in measurements of 47 parallel-sided implants placed in maxillary bone of nine patients. Since RFA measures stability as a function of stiffness, the results indicate that the insertion torque technique can be used to measure bone density and to predict primary stability as measured with RFA. There was no correlation between total mean IT and bone density as assessed using the Lekholm and Zarb index (1985). This may reflect the subjective nature of the latter method which is purely based on the surgeon’s perception of bone density during drilling of the implant site.

It is concluded that the Neoss implant design develops a continuously increasing insertion torque during placement as expected for a tapered implant design. This indicates that the total implant length is involved in lateral compression, providing firm primary stability for the implant. Moreover, there is a correlation between IT and RFA measurements.

REFERENCES


Stress Evaluation of Dental Implant Wall Thickness using Numerical Techniques

Rudi C. van Staden¹, Hong Guan¹, Yew-Chaye Loo¹, Newell W. Johnson², Neil Meredith³,

¹Griffith School of Engineering, Griffith University Gold Coast Campus, Australia;
²School of Dentistry and Oral Health, Griffith University Gold Coast Campus, Australia;
³Neoss Ltd, Harrogate, United Kingdom

Using the Finite Element Method, four implant diameters were evaluated for their effect on the stress distribution at the implant wall. A two-dimensional model of the implant and mandibular bone used triangular and quadrilateral plane strain elements to compute the von Mises stress in the bone with varied masticatory forces and abutment screw preloads. The masticatory force is found to be more influential than abutment screw preload, and implant wall thickness significantly influences the stress magnitude within the implant.

INTRODUCTION

Development of an ideal substitute for missing teeth has been a major aim of dental practitioners for millennia (Irish 2004). Dental implants are biocompatible threaded titanium ‘fixtures’ surgically inserted into the mandible or maxilla to replace missing teeth. The establishment of a good biomechanical link between implant and jawbone is called osseointegration (Branemark et al. 1969, 1977). The success of osseointegration depends on many factors including: medical status of the patient, smoking habits, bone quality, bone grafting, disturbance, sensation or its disruption, haematoma, haemorrhage, tooth necrosis, irradiation therapy, parafunctions, operator experience, degree of surgical trauma, bacterial contamination, lack of preoperative antibiotics, immediate loading, nonsubmerged procedure and implant surface characteristics (Esposito et al. 1998). Excessive surgical trauma and impaired healing ability, premature loading and infection are likely to be the most common causes for early implant losses, whereas progressive chronic marginal infection (peri-implantitis) and overload in conjunction with host characteristics are the major reasons for delayed failures (Esposito et al. 1998). For both early and late implant failures, loading is considered an important factor (Geng et al. 2001). The distribution and magnitude of stresses within the implant are influenced by the implant dimensions, as documented by (Huang et al. 2005, Capodiferro et al. 2006, Winklere et al. 2003, Naert et al. 1992, Tolman and Laney 1992). Catastrophic mechanical failure of the implant may occur by implant fatigue (Huang et al. 2005, Capodiferro et al. 2006), implant fractures, veneering resin/ceramic fractures or other mechanical retention failures (Winklere et al. 2003, Naert et al. 1992, Tolman and Laney 1992). From an engineering perspective, an important criterion in designing an implant is to have a geometry that can minimize mechanical failure caused by an extensive range of loading. As part of the implantation process, the torque is applied to the abutment screw causing an equivalent preload or clamping force between the abutment and implant. This is to ensure that the various implant components are perfectly attached to each other. However, to date no published research appears to have investigated the influence of masticatory forces and abutment screw preloading on stresses in implants of various diameters. Therefore, the aim of this study is to evaluate the stress within an immediately loaded implant under a range of masticatory forces, abutment preloads and implant diameters.
MATERIALS AND METHODS

Modelling and simulation were performed using the Strand7 (2004) Finite Element Analysis (FEA) System. First the implant and bone geometry were defined, then material properties for the implant and various bone components were assigned in terms of Young’s modulus, Poisson’s ratio and density. The loading and restraint conditions were applied and the effect on the stress profile within the implant was evaluated.

Modelling

A two-dimensional (2D) representation of the implant and mandibular bone was analysed because this was considered to be equally accurate and more efficient in computation time, as three-dimensional (3D) modelling. Data were acquired for the bone dimensions by CT scanning. Two different types of bone, cancellous and cortical, were distinguished and the boundaries were identified in order to assign different material properties within the finite element model. Figure 1 shows an implant and mandible section with the loading and restraint conditions. Also shown in the figure are the parameters investigated in this study.

The implant is based on that used by Neoss (2006), it is conical with 2 degrees of taper and a helical thread. For the finite element model with \( D = 4.5 \text{mm}, \quad L = 11 \text{mm}, \quad T_{\text{cor}} = 1.2 \text{mm} \), 3314 plate elements and 3655 nodal points were used for the implant, 3804 elements and 4079 nodes for cancellous bone, and 1216 elements and 1453 nodes for cortical bone.

The effect on stress in the implant wall under different mastication forces (\( F_m \)) and abutment preload (\( F_p \)) was evaluated. For all the possible parametric combinations, the von Mises stress along the line \( VV \) in the implant wall was investigated and is believed to play a crucial role in the probability of implant fracture.

As indicated in Figure 1 the von Mises stresses along the lines \( VV1-2, \ VV2-3 \) and \( VV3-4 \) are reported for all possible combinations of loading and diameters. The relative locations of these lines are also detailed in the figure. Each line is identified by its start and end points, so, for example, line \( VV1-2 \) begins at \( VV1 \) and ends at \( VV2 \). These locations were suggested by clinicians to be prone to micro fractures.

The range of implant diameters (3.5, 4.0, 4.5 and 5.5mm) is shown in Figure 2. Note that the cortical bone is constrained distally to represent normal function of the mandible.

Figure 1. Finite element model of implant, components, implant/bone interface and bone

Figure 2. Finite element model showing different implant diameters
**Plane strain elements**

The model represents a cross-sectional slice from the mandible (Figure 3). The “arc length” of the mandible is comparable to the width and depth of the slice. When the slice is subjected to in-plane (x-y) masticatory forces (FM), it is restrained from deforming out-of-plane (in the z-axis) and it was assumed that all strains are confined in the z-axis. To accurately represent the mechanical behaviour of the bone and implant, 3-node triangular (Tri3) and 4-node quadrilateral (Quad4) plane strain elements were therefore used for the construction of the finite element models. Note that for plane strain elements each node has a complete set of spatial degrees of freedom including u and v. The constraints meant that nodes could only translate in the x- (u) and y- (v) directions.

**Material properties**

The material properties adopted were specified in terms of Young's modulus, Poisson's ratio and density for the implant and all associated components (Table 1). All materials were assumed to exhibit linear, homogeneous elastic behaviour.

**Loading conditions**

Loading conditions included the masticatory force, FM, applied to the occlusal surface of the crown set at 200, 500 and 1000N, as shown in Figure 1, and applied at 45° inclination in the x y plane (refer to Figure 1). A preload, FP, was applied to the abutment screw of magnitude 201.93, 587.44 or 1009.67 N through the use of temperature sensitive elements. The technique for applying the masticatory forces to the crown, and the preload applied to the abutment screw, are discussed below.

**Masticatory force, FM**

During normal function the crown is subjected to oblique loads applied to the occlusal surface of the crown. These loads are a result of normal masticatory function. The theoretical study by Choi et al. (2005) suggests that this loading condition can be considered to represent all applied loads. The implant was restrained in the x, y and z directions within the jawbone, assuming complete osseointegration at the bone and implant interface. Figure 1 shows the positions of the applied loading and restraints.

<table>
<thead>
<tr>
<th>Component</th>
<th>Description</th>
<th>Young’s Modulus, E (GPa)</th>
<th>Poissons ratio, v</th>
<th>Density, ρ (g/cm³)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Implant, abutment, washer</td>
<td>Titanium (grade 4)</td>
<td>105.00</td>
<td>0.37</td>
<td>4.51</td>
</tr>
<tr>
<td>Abutment screw</td>
<td>Gold (precision alloy)</td>
<td>93.00</td>
<td>0.30</td>
<td>16.30</td>
</tr>
<tr>
<td>Crown</td>
<td>Zirconia (Y-TZP)</td>
<td>172.00</td>
<td>0.33</td>
<td>6.05</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td></td>
<td>1.00</td>
<td>0.30</td>
<td>0.74</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>Cortical thickness = 1.2mm</td>
<td>13.70</td>
<td>0.30</td>
<td>2.19</td>
</tr>
</tbody>
</table>
Preload, $F_p$

The torque applied to the abutment screw causes the preload or clamping forces between the implant and abutment. The procedure for calculating $F_p$ and applying a temperature sensitive element, functioning throughout the abutment screw developing the preload (or torque), is described below. Note that for the purpose of this article only the Neoss system calculations for $F_p$ and temperature sensitive elements are shown as an example.

A dental implant system typically consists of a crown, abutment and abutment screw. The abutment screw is screwed with a manual screwdriver (Figure 4) into the internal thread of the implant. Finally, the crown is placed on to the abutment using cement at the crown and abutment interface.

The nature of the forces clamping implant components together, and how they are generated and sustained, are not comprehensively covered in the literature. Preload was considered in finite element modelling by Haack et al. 1995, Lang et al. 2003 and Byrne et al. 2006. These studies were either based on complicated 3D modelling or did not specify the techniques used for replicating $F_p$. For the purpose of this study the calculation used to determine the preload is based on findings by Dekker (1995).

The relationship between the torque applied to the abutment screw and the preload achieved was expressed by Dekker (1995) as:

$$T_p = F_p \left( \frac{i}{2\pi} + \frac{\mu_i r_t}{\cos \beta} + \mu_{it} r_t \right)$$  

using the conditions listed in Table 2;

Calculations of the preload are shown in equation 2 below, using information derived in equation 1.

$$F_p = \frac{320}{\left( \frac{0.4}{2\pi} + \frac{(0.8 \times 0.8725)}{\cos (8.2)} + (0.9 \times 1.125) \right)}$$

Rearranging for $F_p$:

$$F_p = \frac{320}{\left( \frac{0.4}{2\pi} + \frac{(0.8 \times 0.8725)}{\cos (8.2)} + (0.9 \times 1.125) \right)}$$

$F_p = 587.44$N

Table 2. Conditions applied in equation 1

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
<th>Magnitude</th>
<th>Reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T_p$</td>
<td>torque applied to the abutment screw (Nmm)</td>
<td>320</td>
<td>Neoss (2006)</td>
</tr>
<tr>
<td>$F_p$</td>
<td>preload created in the abutment screw (N)</td>
<td>Unknown factor</td>
<td></td>
</tr>
<tr>
<td>$i$</td>
<td>the pitch of the abutment screw threads (mm)</td>
<td>0.40</td>
<td>Neoss (2006)</td>
</tr>
<tr>
<td>$\mu_i$</td>
<td>the coefficient of friction between abutment screw thread surfaces and internal implant screw thread surfaces (dimensionless)</td>
<td>0.26</td>
<td>Lang et al. (2003)</td>
</tr>
<tr>
<td>$r_t$</td>
<td>the effective contact radius between the inner implant thread and the abutment screw threads (mm)</td>
<td>((r_t + r_3) / 2 = (0.99 + 0.755)/2 = 1.745 / 2 = 0.87)</td>
<td>Neoss (2006)</td>
</tr>
<tr>
<td>$\mu_{it}$</td>
<td>the half-angle of the threads (degree)</td>
<td>28.72°</td>
<td></td>
</tr>
<tr>
<td>$r_t$</td>
<td>the effective radius of contact between the abutment and implant surface (mm)</td>
<td>((r_t + r_3) / 2 = (1.275 + 0.95)/2 = 2.225 / 2 = 1.11)</td>
<td>Neoss (2006)</td>
</tr>
</tbody>
</table>
The preload clamping the abutment to the implant is transferred through two interfaces. The first interface (SA1) is between the abutment and abutment screw and the second (SA2), between the abutment screw threads and the inner threads of the implant (Figure 5). The calculated preload, \( F_P \), is assumed to act equally, as a pressure, \( q \), across the first and second interfaces. Due to equilibrium, only the pressure \( q \), acting on SA1, is considered in this study.

\[
SA_1 = (\pi \times r_1^2) - (\pi \times r_2^2) = (\pi \times 1.275^2) - (\pi \times 0.952^2)
SA_1 = 2.27\text{mm}^2
\]

The pressure acting on \( SA_1 \), when \( F_P = 587.77\text{N} \), was calculated as follows:

\[
q = \frac{F_P}{SA_1} = \frac{587.4}{2.2748984} \ldots \ldots \ldots \ldots \ldots (3)
\]

\( q = 258.22\text{ N/mm}^2 \)

The clamping pressure, \( q \), is a result of the applied torque and is a means of replicating the 3D torque in a 2D manner. For the present study, \( q \), is calculated as above for the applied torques of 110, 320 and 550Nmm (Table 3).

In the analysis a negative temperature (-10 Kelvin, K) is applied to all the nodal points within the abutment screw, causing each element to shrink. A trial and error process is applied to determine the temperature coefficient, \( C \), that can yield an equivalent \( q \). The corresponding \( C \) is also presented in Table 3.

Table 3. Abutment screw torque \((T)\), preload \(F_P\), and surface pressure \(q\)

<table>
<thead>
<tr>
<th>( T ) (Nmm)</th>
<th>( F_P ) (N)</th>
<th>( q ) (N/mm(^2))</th>
<th>( C_{3.0} ) (( \times 10^{-12} ))</th>
<th>( C_{4.0} ) (( \times 10^{-12} ))</th>
<th>( C_{4.5} ) (( \times 10^{-12} ))</th>
<th>( C_{5.0} ) (( \times 10^{-12} ))</th>
</tr>
</thead>
<tbody>
<tr>
<td>110.00</td>
<td>201.93</td>
<td>88.76</td>
<td>-2.62</td>
<td>-2.54</td>
<td>-3.52</td>
<td>-4.14</td>
</tr>
<tr>
<td>320.00</td>
<td>587.44</td>
<td>258.22</td>
<td>-7.61</td>
<td>-7.39</td>
<td>-10.20</td>
<td>-12.71</td>
</tr>
<tr>
<td>550.00</td>
<td>1009.67</td>
<td>443.83</td>
<td>-13.08</td>
<td>-12.67</td>
<td>-17.64</td>
<td>-20.78</td>
</tr>
</tbody>
</table>
RESULTS

The distribution of von Mises stresses in the implant is discussed for all combinations of masticatory and preload forces. Each parameter is discussed in a separate section for all four implant diameters. The von Mises stresses are reported between locations VV1-2 (0 - 1.51mm in length), VV2-3 (1.51 - 2.24mm) and NN3-4 (2.24 - 3.77mm), as shown in Figure 1.

Masticatory Force, $F_M$

The distributions of von Mises stresses along the lines VV1-2, VV2-3, and VV3-4 for all values of $F_M$ are shown in Figure 6. Note that the preload, $F_P$, is set at its medium value, i.e. 587.44N.

In general, when the applied masticatory force, $F_M$, is increased, the von Mises stresses also increase proportionally, because the system being analysed is linear elastic. As expected the 3.5mm implant shows higher stresses than all other diameters (refer to Figures 6 a and b). The 3.5mm also indicates stress peaks along the lines VV1-2 and VV2-3 where all other parameters only have elevated stress peaks along the line VV1-2 (Figures 6 a, c, e and g). This is because the implant wall thickness for the 3.5mm implant is significantly reduced in the region corresponding to VV2-3, hence causing stress concentration. The 4.0 and 4.5mm diameter implants have similar stress distribution characteristics but the stresses are lower in magnitude at VV2-3, VV3-4 and VV4-5 than with the 3.5mm implant because of the greater wall thickness (Figures 6 d and f). The 5.5mm implant displays greatly reduced stresses at all locations (Figures 6 g and h), with peak stresses occurring close to point VV1.

Preload Force, $F_P$

To investigate the effect of different preload $F_P$, $F_M$ is kept as a constant and its medium value, i.e. 500N is considered herein. The distributions of von Mises stresses along the lines VV1-2, VV2-3, and VV3-4 for all values of $F_P$ are shown in Figure 7. Similar stress distribution characteristics were found when varying $F_P$ as with $F_M$. Note that when $F_P$ increases, the von Mises stresses also increase. However, the increase is not proportional to the increase of $F_P$. This is because $F_P$ as an internal force, is a function of the abutment screw and implant diameters. This suggests that failure of the crown is more likely to be caused by $F_M$.

DISCUSSION

FEA has been used extensively to predict the biomechanical and mechanical performance of implant designs as well as the effect of clinical factors on the success of implantation (DeTolla et al. 2000, Geng et al. 2001). The principal difficulty in simulating the mechanical behaviour of implants is the modelling of the living human bone tissue and its response to applied mechanical forces (van Staden et al. 2006). A few studies have considered the influence of such factors as the torque (Lang et al. 2003) with which the abutment is connected, and the effect of masticatory forces on the probability of implant failure or loosening (Byrne et al. 2006).

This paper considers various combinations of loading parameters and their influence on the stress produced within implants of diameters 3.5, 4.0, 4.5 and 5.5mm. As expected, increased masticatory forces lead to greater stresses within the implant. The preload applied to the abutment has less influence on the stresses than the masticatory forces. The 3.5mm implant shows higher stresses than all other diameters and indicates stress peaks along the lines VV1-2 and VV2-3 where all other parameters only have elevated stress peaks along the line VV1-2. The 4.0 and 4.5mm diameter implants have similar stress distribution characteristics and the 5.5mm implant displays greatly reduced stresses at all locations, with peak stresses occurring close to point VV1.

CONCLUSION

Overall, it was found that the masticatory force is more influential on implant stresses than the abutment screw preload. As expected the implant wall thickness significantly influences the stress magnitude within the implant. Note also that when the wall thickness is decreased (especially for the 3.5mm) stress concentration occurs at the internal and external threads as well as at sharp corners of the implant wall.

Characteristically the stress showed an increase at the top of the implant thread and the top of the implant (line VV1-2). It should be noted that this research was a pilot study aimed at offering an initial understanding of the complicated stress distribution characteristics due to the various parameters. Other parameters which may be evaluated for their influence on implant stress profiles include the implant taper, pitch and design of
Figure 6. Stress characteristics when varying $F_w$.

- a) Stress profile
- b) Stress contour
- c) Stress profile
- d) Stress contour
- e) Stress profile
- f) Stress contour
- g) Stress profile
- h) Stress contour
Figure 7. Stress characteristics when varying $F_P$. 

$a)$ Stress profile

$b)$ Stress contour

$c)$ Stress profile

$d)$ Stress contour

$e)$ Stress profile

$f)$ Stress contour

$g)$ Stress profile

$h)$ Stress contour
Implant thread, implant neck offset, different stages of bone remodelling and implant orientation within the bone. It is important that clinicians understand the methodology, applications, and limitations of FEA in implant dentistry, and become more confident in interpreting results from FEA studies to clinical situations.

ACKNOWLEDGEMENTS

This work was made possible by the collaborative support from Griffith’s School of Engineering and School of Dentistry and Oral Health. A special thank you goes to Messer John Divitini and Fredrik Engman from Neoss Limited for their continual contribution.

REFERENCES


Comparative Analysis of Two Implant-Crown Connection Systems - A Finite Element Study

Rudi C. van Staden¹, Hong Guan¹, Yew-Chaye Loo¹, Newell W. Johnson², Neil Meredith³,

¹Griffith School of Engineering, Griffith University Gold Coast Campus, Australia;
²School of Dentistry and Oral Health, Griffith University Gold Coast Campus, Australia;
³Neoss Ltd, Harrogate, United Kingdom

The internal and external-hex connections of the Neoss and 3i implant systems were compared in a three-dimensional Finite Element Analysis. Chewing forces of 200, 500 and 1000N and abutment screw preloads of 110, 320 and 550Nmm were studied. The connection type strongly influences the stress profile within the crown, with the external-hex connection exhibiting greater stresses than the internal-hex.

INTRODUCTION

Dental implants are a well accepted treatment for partially or totally edentulous subjects. Innovations through research have led to advancements in surgical and restorative techniques, improved surface features and restorative components. Dental implants typically use either internal-hex or external-hex connections with the crown (Figure 1 a). Although both connections are extensively used clinically, distinctly different stress distributions are produced within the crown. Clinicians have reported implant components linked to mechanical failure of crown and implant (Maeda et al. 2006, Merz et al. 2000, Khraisat et al. 2002). Two major factors may be implicated in crown and implant failure. These are;

- over tightening of the abutment screw leading to failure of the crown for internal-hex and external-hex connection type implant systems.
- excessive masticatory loads transferred from the occlusal plane of the crown to a stress concentration at the interface between the abutment and implant body.

Theoretical techniques such as the Finite Element Method (FEM) can be used to evaluate mechanical factors that could affect implant performance and success (Capodiferro et al. 2006, Gehrke et al. 2006, Huang et al. 2005, Khraisat et al. 2005). This study was undertaken using Finite Element Analysis (FEA) to aid understanding of the stresses in both internal-hex and external-hex implant systems under different loading conditions.

Studies by Maeda et al. (2006), Khraisat et al. (2002) and Merz et al. (2000) have all considered the stress within the abutment screw but disregarded the stress within the crown. To date no published research appears to have investigated the stress profile in the crown due to an internal-hex or external-hex connection. Ultimately, the outcome of this study will help dental practitioners to identify locations within the implant system susceptible to stress concentrations.

MATERIALS AND METHODS

The modelling and simulation herein are performed using the Strand7 FEA System (2004). The first step of the modelling is to define the geometry of the implant system. Then the material behaviour of the model is specified in terms of Young’s modulus, Poisson’s ratio and density for all components. The appropriate loading and restraint conditions are applied and the individual parameters and their contribution to the stress profile within the crown and implant is evaluated.

Modelling

Geometry for the implant systems were obtained from the manufacturers (Figure 1a). Section AA is set at a location where maximum compressive stresses occur within the crown, positioned at 30° from the x-axis towards the negative z-axis.

The parameters investigated are shown in Figure 1b. The implant is conical with 2 degrees of taper, a helical thread, outside diameter of 4.5mm and length of
11mm. Half the implant is modelled and symmetrical constraints are applied along the plane of symmetry as indicated (Figure 1b). For the Neoss (2006) and 3i (2006) finite element models, the total numbers of elements are respectively 13464 and 30420 for the implant, 3564 and 9108 for the abutment, 17424 and 25956 for the abutment screw, and 38484 and 47052 for the crown. The total number of nodal points for the entire Neoss model are 122688 and 82547 for the 3i.

The main focus of this study is to examine the stress characteristics within the crown and next to the crown-abutment interface. Therefore an assumption is made to restrain the outer edge of the implant (Figure 1b) when the mandibular bone is not included in the finite element model. Note that these loading and restraint conditions are the same for both internal and external-hex systems.
Stress reporting
The von Mises stresses along the lines NN (NN1-2, NN2-3 and NN3-4) and II (II1-2, II2-3, II3-4, II4-5, II5-6 and II6-7) for both systems are reported for all possible combinations of loading (Figure 1c). The relative locations of these lines are also detailed in the figure, which are identified by their start and end points. So, for example, the line II1-2 begins at II1 and ends at II2. The von Mises stresses along the lines NN and II, in the crown, are believed to play a crucial role in the probability of crown fracture. On Section AA lines NN and II were chosen because the greatest stresses due to the masticatory loading (compressive is prominent over tensile) occur on this plane.

Material properties
The material properties in the model are specified in terms of Young’s modulus, Poisson’s ratio and density for the implant and all associated components (Table 1). All materials are assumed to exhibit linear, elastic and homogeneous behaviour.

Loading conditions
Masticatory force, \( F_{m} \), was applied to the occlusal surface of the crown at 100, 250 or 500N, inclined at 45° to the x and y-axes (Figure 1b). The preload, \( F_{p} \), of 100.97, 293.72 or 504.84N is applied to the abutment screw through the use of temperature sensitive elements (Figure 1b). Note that \( F_{m} \) and \( F_{p} \) are set to half of the total magnitude because only half of the implant system is modelled. Therefore the total \( F_{m} \) modelled is 200, 500, 1000N and \( F_{p} \) is 201.93, 587.44, 1009.67N.

The manner of modelling the masticatory forces and the preload applied to the abutment screw was described by van Staden et al. (2008). For the purposes of this study both the abutment screw preload, \( F_{p} \), and the surface area between abutment and abutment screw, \( A_{s} \), are halved when compared to those used by van Staden et al. (2008) due to the modelling assumption (half model). Calculations for the abutment screw surface pressure, \( q \), confer identical results to those found by van Staden et al. (2008).

For the present study a negative temperature (-10 Kelvin, K) was applied to all the nodal points within the abutment screw, causing each element to shrink. A trial and error process was applied to determine the temperature coefficient, \( C \), for the Neoss and 3i systems (i.e. \( C_{Neoss} \) and \( C_{3i} \) ) that can yield an equivalent \( q \). It was found that when \( F_{p} = 201.93, 587.44 \) and 1009.67N then \( C_{Neoss} = -3.51×10^{-4}, -9.28×10^{-4} \) and \(-15.60×10^{-4}/K\), and \( C_{3i} = -0.98×10^{-4}, -1.80×10^{-4} \) and \(-2.68×10^{-4}/K\), respectively.

RESULTS
Zirconia, typically used as a dielectric material, has proven adequate for application in dentistry. With its white appearance and high Young’s modulus it is ideal for use in sub frames for dental restorations such as crowns and bridges, which can then be veneered with conventional feldspathic porcelain. Zirconia has a fracture strength greater than titanium; therefore it may be considered as a high strength material. However with masticatory and preload forces the compressive strength of 2.1 GPa (Curtis et al. (2005)) can easily be exceeded; especially for implant systems with external-hex connections, as confirmed during this study.

The distribution of von Mises stresses in the crown is discussed for all parametric combinations of masticatory and preload forces. Each parameter is discussed in a separate section. For the Neoss system, the von Mises stresses are reported between locations NN1-2 (0-1.76mm in length), NN2-3 (1.76-1.87mm) and NN3-4 (1.87-3.96mm). For the 3i system the von Mises stresses are reported between locations II1-2 (0-2.38mm), II2-3 (2.38-2.78mm), II3-4 (2.78-3.67mm), II4-5 (3.67-4.06mm), II5-6 (4.06-4.65mm) and II6-7 (4.65-5.27mm), as shown in Figure 1c.

Masticatory Force, \( F_{m} \)

The distributions of von Mises stresses along the lines NN and II for all values of \( F_{m} \) are shown in Figure 2. Note that the preload, \( F_{p} \), is set at its medium value, i.e. 587.44N.

In general, when the applied masticatory force, \( F_{m} \), is increased, the von Mises stresses also increase propor-

---

Table 1. Material properties

<table>
<thead>
<tr>
<th>Component</th>
<th>Description</th>
<th>Young’s Modulus, E (GPa)</th>
<th>Poisson’s ratio, v</th>
<th>Density, ( \rho ) (g/cm^3)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Implant, abutment, washer</td>
<td>Titanium (grade 4)</td>
<td>105.00</td>
<td>0.37</td>
<td>4.51</td>
</tr>
<tr>
<td>Abutment screw</td>
<td>Gold (precision alloy)</td>
<td>93.00</td>
<td>0.30</td>
<td>16.30</td>
</tr>
</tbody>
</table>
tionally, because the system being analysed is linear elastic. When \( F_M \) increases the stress along the line \( NN \) increases showing two peaks along the line \( NN_{3-4} \) (refer to Figure 2a). The larger of these two peaks occurs at a distance of ±3.8mm in length from \( NN_1 \). This stress peak (as can be identified in Figure 2 b) is caused by a sharp corner and sudden change in section at this point.

Elevated stress concentrations are identified at the beginning of the line \( II_{3-4} \) (Figures 2 c and d). This stress peak, as can be identified in Figure 2 c, is caused by a sharp corner at this point. For the 3i system the volume of the crown exceeds that of the Neoss system, thereby suggesting that the 3i crown may show greater resistance to the applied masticatory forces. However, even though the Neoss crown has a thinner wall along the line \( NN_{3-4} \), reduced stresses are still evident due to the abutment’s high Young’s modulus. Overall, the design differences between the Neoss and 3i systems result in higher stresses in the 3i system when \( F_M \) is increased.

**Preload Force, \( F_P \)**

To investigate the effect of different preload \( F_P \), \( F_M \) is kept as a constant and its medium value, i.e. 500N is considered herein. The distributions of von Mises stresses along the lines \( NN \) and \( II \) for all values of \( F_P \) are shown in Figure 3.

As found in Section 3.1 for \( F_M \) when \( F_P \) increases the stresses calculated along the line \( NN \) increase, showing two peaks along the line \( NN_{3-4} \) (refer to Figures 3 a and b). Also, as found for \( F_M \), elevated stress peaks are identified at the beginning of the line \( II_{3-4} \) (Figures 3 c and d). Overall, all values of \( F_P \) cause greater stresses along lines \( NN \) and \( II \), than do varying values of \( F_M \).
DISCUSSION

FEA has been used extensively to predict the biomechanical performance of the jawbone surrounding a dental implant (DeTolla et al. 2000, Geng et al. 2001). Previous research considered the influence of the implant dimensions and the bone-implant bond on the stress in the surrounding bone. However, to date no research has been conducted to evaluate the stress produced by different implant to crown connections (i.e. internal-hex and external-hex).

The analysis completed in this paper uses the FEM to replicate internal-hex and external-hex connections for loading parameters of $F_M$ and $F_P$. As shown in Table 2, two stress peaks were revealed along the lines NN and II at locations 3.76 and 2.89mm from the top. The stress values shown were calculated with the other parameter (i.e. $F_M$ or $F_P$) set to its average. The mastication force $F_M$ is applied on the occlusal surface of the crown, evenly distributed along 378 nodal locations (Figure 1 b), and orientated at 45° in the x-y plane. This induces compressive stresses in the right hand side of the crown and tensile in the left. Varying $F_M$ from 200 to 1000N for the internal-hex and external-hex systems results in a change in von Mises stress of 545.64 (818.47-272.82MPa) and 698.09MPa (1047.14-349.05MPa) respectively. The geometrical design of the external-hex system tends to induce stress concentrations, located

Table 2. von Mises stress (MPa) in the crown (location of stress reporting in brackets)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>$F_M$ (N)</th>
<th>$F_P$ (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Line</td>
<td>200</td>
<td>500</td>
</tr>
<tr>
<td>NN (3.76mm)</td>
<td>272.82</td>
<td>545.64</td>
</tr>
<tr>
<td>II (2.89mm)</td>
<td>349.05</td>
<td>698.09</td>
</tr>
</tbody>
</table>
2.89mm from the apex in this study. For this system, a stress concentration at this point is also induced by $F_p$, increasing the compressive stresses on the right hand side of the crown. Increasing $F_p$ from its minimum to maximum values, for the external-hex system, increases the stress by 485.46MPa (951.67-466.21MPa).

The internal-hex system has reduced stress concentrations, demonstrating that this design is less susceptible to stress concentrations within the crown. However, because of the transfer of the preload through the abutment screw to abutment contact, changing $F_p$ is more influential on this hex system than $F_{oc}$. Overall $F_p$ is more influential on the stress within the crown for the external-hex system and $F_p$ is more influential on the internal-hex system.

**CONCLUSION**

This research is a pilot study aimed at offering an initial understanding of the stress distribution characteristics in the crown under different loading conditions. Realistic geometries, material properties, loading and support conditions for the implant system were used.

The geometrical design of the external-hex system tends to induce stress concentrations in the crown at a distance of 2.89mm from the apex. At this location $F_p$ also affects the stresses, so that the compressive stresses on the right hand side of the crown are increased. The internal-hex system has reduced stress concentrations in the crown. However, because the preload is transferred through the abutment screw to the abutment contact, changing $F_p$ has greater effect on this hex system than $F_{oc}$. Overall $F_p$ is more influential on the stress within the crown for the external-hex system and $F_p$ is more influential on the internal-hex system.

Future recommendations include the evaluation of other implant parameters such as the implant wall thickness and thread design. Ultimately, all implant components can be understood in terms of their influence on the stress produced within the implant itself.

**ACKNOWLEDGEMENTS**

This work was made possible by the collaborative support from Griffith’s School of Engineering and School of Dentistry and Oral Health. A special thank you goes to Messires John Divitini and Ian Kitchingham from Neoss Limited for their continual contribution.

**REFERENCES**


Comparison of early bacterial colonization of PEEK and titanium healing abutments using real-time PCR

Stefano Volpe¹, Damiano Verrocchi², Peter Andersson², Jan Gottlow³, Lars Senerby⁵
¹Private Practice, Rome, Italy
²Private Practice, Fiera di Primiero and Feltre, Italy
³Dept Biomaterials, Inst Clinical Sciences, Sahlgrenska Academy, Gothenburg University, Sweden

This study found no differences in total bacterial load at PEEK or titanium Neoss healing abutment surfaces or their adjacent peri-implant pockets/sulci. The number of periodontal pathogens was low and there was no significant difference between the abutment materials. The findings support the use of PEEK as abutment material in dental implant care.

INTRODUCTION

PEEK (polyetheretherketone) is a synthetic polymer with high biomechanical strength and inert chemical properties, which make it attractive for use in industrial and medical applications (for review see Kurtz & Devine 2007). Healing abutments made of commercially pure titanium have been the “gold standard” in dental implant care for many years. Wound healing studies have documented the formation of a healthy mucosal barrier adhering to the titanium abutment surface after abutment connection (e.g. Berglundh et al 1991). Yet, it has been shown that peri-implant pockets are colonized by bacteria, including periodontal pathogens, already within 2 weeks after abutment connection (Koka et al 1993, Quirynen et al 2006). The Neoss implant system offers both titanium and PEEK healing abutments.

The aim of the present study was to evaluate the biocompatibility of PEEK abutments to that of titanium abutments by comparing the bacterial colonization of the abutment surfaces and the surrounding peri-implant pocket/sulcus using real-time PCR.

MATERIALS AND METHODS

Patient and sampling sites
Fourteen (14) partially edentulous patients (9 women and 5 men with a mean age of 58 years) with need of implant treatment volunteered for the study. All patients were of good general health and had not received any antibiotic therapy within the previous 3 months. Every patient received 2 submerged Neoss implants in one edentulous area. Second stage surgery was performed after 3-6 months of healing. Each patient received one titanium and one PEEK healing abutment. A chlorhexidine rinse regimen was instituted during the first postoperative week. Sutures were removed after 7 days.

Bacterial sampling was performed two weeks after abutment connection using a commercial test system (Meridol® Perio Diagnostics, GABA International, Münchenstein, Switzerland). Each test kit contains 4 paperpoints for sampling. All samples were taken at the distal surface of the abutments in order to stay as far as possible away from the neighbouring anterior teeth. Two paperpoints were used at each titanium and PEEK abutment site. The first paperpoint was positioned in close contact with the abutment surface at the mucosal margin. The second paperpoint was placed in the peri-abutment sulcus/pocket area after removal of the abutment. Each paperpoint was held in position for 10 seconds and immediately thereafter placed in a test tube (accompanying the test kit). The samples were sent to a specialized microbiological laboratory (Carpagen GmbH, Münster, Germany).
Real-time PCR

Real-time polymerase chain reaction (PCR) by the Meridol® Perio Diagnostics (GABA International, Münchenstein, Switzerland) detects and quantifies six periodontal pathogens (\textit{A. actinomycetemcomitans, P. nucleatum ssp., P. gingivalis, P. intermedia, T. forsythensis} and \textit{T. denticola}) and the total bacterial load.

Statistical analysis

The Wilcoxon signed rank test was used to calculate the significance of the differences found between numbers of bacteria at titanium and PEEK abutment sites.

RESULTS

The study involved 28 healing abutments (14 PEEK and 14 titanium abutments) in 14 patients. At the day of bacterial sampling all sites showed clinically healthy tissue conditions (Fig. 1 and 2).

The total bacterial load at the mucosal margin was on the average $13 \times 10^6$ (median $5 \times 10^6$) at titanium abutments and $14 \times 10^6$ (median $7 \times 10^6$) at PEEK abutments (table 1). The corresponding value for the peri-implant pocket/sulcus was $9 \times 10^6$ (median $2 \times 10^6$) at titanium abutments and $3 \times 10^6$ (median $1 \times 10^6$) at PEEK abutments (table 2). The differences were not statistically significant.

The total number of the 6 periodontal pathogens detectable by the PCR test was on the average $18 \times 10^3$ (median $1 \times 10^4$) at titanium abutments and $14 \times 10^3$ (median $2 \times 10^3$) at PEEK abutments (table 3). The corresponding value for the peri-implant pocket/sulcus was $23 \times 10^3$ (median $2 \times 10^3$) at titanium abutments and $75 \times 10^3$ (median $8 \times 10^3$) at PEEK abutments (table 4). No sample showed $\geq 1 \times 10^6$ of periodontal pathogens. The differences between titanium and PEEK abutments were not statistically significant.

DISCUSSION

The Neoss PEEK abutments showed similar amounts of bacterial adhesion as the Neoss titanium abutments when evaluated using real-time PCR. This method is very sensitive since both living and dead bacteria are detected, which makes it more accurate than cultivation. The detection limit for each periodontal pathogen is as low as 100 bacteria (Jervøe-Storm et al 2005).

Hultin and coworkers (2002) described microbiological findings and host response in partially edentulous patients with peri-implantitis. DNA-probe analysis sensitive for \textit{A. actinomycetemcomitans, P. gingivalis, P. intermedia, T. forsythensis} and \textit{T. denticola} found presence of periodontal pathogens at healthy as well as at diseased implant sites. However, only around implants with peri-implantitis were all 5 species recovered in amounts $\geq 1 \times 10^6$ of the target bacterial cells in each sample. No sample in the present study showed $\geq 1 \times 10^6$ of periodontal pathogens indicating healthy conditions at the tested sites.
The findings of the present study support continued investigation of PEEK as an abutment material in dental implant care.

ACKNOWLEDGEMENT
The authors want to thank Dr. Emanuele Leoncini, Medical Faculty, University La Sapienza, Rome, Italy, for the statistical analyses.

REFERENCES


### Table 2. Total bacterial load - sulcus

<table>
<thead>
<tr>
<th>Patient</th>
<th>Titanium</th>
<th>PEEK</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>4 400 000</td>
<td>220 000</td>
</tr>
<tr>
<td>2</td>
<td>27 000 000</td>
<td>7 800 000</td>
</tr>
<tr>
<td>3</td>
<td>1 500 000</td>
<td>16 000 000</td>
</tr>
<tr>
<td>4</td>
<td>10 000 000</td>
<td>3 000 000</td>
</tr>
<tr>
<td>5</td>
<td>39 000 000</td>
<td>550 000</td>
</tr>
<tr>
<td>6</td>
<td>370 000</td>
<td>1 300 000</td>
</tr>
<tr>
<td>7</td>
<td>650 000</td>
<td>310 000</td>
</tr>
<tr>
<td>8</td>
<td>1 600 000</td>
<td>430 000</td>
</tr>
<tr>
<td>9</td>
<td>1 400 000</td>
<td>2 600 000</td>
</tr>
<tr>
<td>10</td>
<td>1 700 000</td>
<td>170 000</td>
</tr>
<tr>
<td>11</td>
<td>1 000 000</td>
<td>6 700 000</td>
</tr>
<tr>
<td>12</td>
<td>1 600 000</td>
<td>190 000</td>
</tr>
<tr>
<td>13</td>
<td>150 000</td>
<td>1 200 000</td>
</tr>
<tr>
<td>14</td>
<td>31 000 000</td>
<td>4 300 000</td>
</tr>
<tr>
<td>Mean</td>
<td>8 669 286</td>
<td>3 197 857</td>
</tr>
<tr>
<td>Median</td>
<td>1 600 000</td>
<td>1 250 000</td>
</tr>
<tr>
<td>Min</td>
<td>150 000</td>
<td>170 000</td>
</tr>
<tr>
<td>Max</td>
<td>39 000 000</td>
<td>16 000 000</td>
</tr>
<tr>
<td>Sum</td>
<td>121 370 000</td>
<td>44 770 000</td>
</tr>
</tbody>
</table>

\[ p = 0.158 \]

### Table 3. Total amount of periodontal pathogens - surface

<table>
<thead>
<tr>
<th>Patient</th>
<th>Titanium</th>
<th>PEEK</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1 300</td>
<td>1 490</td>
</tr>
<tr>
<td>2</td>
<td>1 450</td>
<td>1 310</td>
</tr>
<tr>
<td>3</td>
<td>177 040</td>
<td>145 600</td>
</tr>
<tr>
<td>4</td>
<td>2 710</td>
<td>7 400</td>
</tr>
<tr>
<td>5</td>
<td>750</td>
<td>750</td>
</tr>
<tr>
<td>6</td>
<td>600</td>
<td>600</td>
</tr>
<tr>
<td>7</td>
<td>1 380</td>
<td>900</td>
</tr>
<tr>
<td>8</td>
<td>920</td>
<td>5 590</td>
</tr>
<tr>
<td>9</td>
<td>600</td>
<td>1 050</td>
</tr>
<tr>
<td>10</td>
<td>750</td>
<td>3 330</td>
</tr>
<tr>
<td>11</td>
<td>1 800</td>
<td>11 790</td>
</tr>
<tr>
<td>12</td>
<td>1 170</td>
<td>1 700</td>
</tr>
<tr>
<td>13</td>
<td>2 100</td>
<td>6 500</td>
</tr>
<tr>
<td>14</td>
<td>53 630</td>
<td>2 410</td>
</tr>
<tr>
<td>Mean</td>
<td>17 586</td>
<td>13 601</td>
</tr>
<tr>
<td>Median</td>
<td>1 340</td>
<td>2 055</td>
</tr>
<tr>
<td>Min</td>
<td>600</td>
<td>600</td>
</tr>
<tr>
<td>Max</td>
<td>177 040</td>
<td>145 600</td>
</tr>
<tr>
<td>Sum</td>
<td>246 200</td>
<td>190 420</td>
</tr>
</tbody>
</table>

\[ p = 0.388 \]

### Table 4. Total amount of periodontal pathogens - sulcus

<table>
<thead>
<tr>
<th>Patient</th>
<th>Titanium</th>
<th>PEEK</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>6 960</td>
<td>6 350</td>
</tr>
<tr>
<td>2</td>
<td>2 000</td>
<td>8 700</td>
</tr>
<tr>
<td>3</td>
<td>15 400</td>
<td>895 550</td>
</tr>
<tr>
<td>4</td>
<td>20 760</td>
<td>25 530</td>
</tr>
<tr>
<td>5</td>
<td>1 340</td>
<td>13 950</td>
</tr>
<tr>
<td>6</td>
<td>600</td>
<td>850</td>
</tr>
<tr>
<td>7</td>
<td>600</td>
<td>600</td>
</tr>
<tr>
<td>8</td>
<td>900</td>
<td>750</td>
</tr>
<tr>
<td>9</td>
<td>4 350</td>
<td>67 750</td>
</tr>
<tr>
<td>10</td>
<td>5 210</td>
<td>600</td>
</tr>
<tr>
<td>11</td>
<td>900</td>
<td>750</td>
</tr>
<tr>
<td>12</td>
<td>1 100</td>
<td>10 220</td>
</tr>
<tr>
<td>13</td>
<td>750</td>
<td>14 130</td>
</tr>
<tr>
<td>14</td>
<td>261 390</td>
<td>1 950</td>
</tr>
<tr>
<td>Mean</td>
<td>23 019</td>
<td>74 834</td>
</tr>
<tr>
<td>Median</td>
<td>1 670</td>
<td>7 525</td>
</tr>
<tr>
<td>Min</td>
<td>600</td>
<td>600</td>
</tr>
<tr>
<td>Max</td>
<td>261 390</td>
<td>895 550</td>
</tr>
<tr>
<td>Sum</td>
<td>322 260</td>
<td>1 047 680</td>
</tr>
</tbody>
</table>

\[ p = 0.133 \]
The possible causes of implant fracture under dynamic loading conditions are discussed with regard to the in-vivo clinical behaviour and in-vitro testing.

INTRODUCTION
Fracture of endosseous dental implants during placement or function will have serious implications for patients. Fracture of implants during insertion may occur as the insertion load exceeds the fracture strength of the implant. Such a failure is most unlikely to be the result of clinical misuse and is most probably due to an error in design or material selection. Errors in manufacturing and flaws in materials may also contribute to failure. A design error is likely to involve a number of components whereas manufacturing and material errors may be limited to a single component or batch of raw material.

Failures at placement are generally noticed clinically at the time. However, should the clinician fail to notice a crack or flaw and the component was then incorporated in a structure then potentially serious bone loss and clinical complications may arise which may not be apparent for some months or even years post restoration.

The other mechanism of failure for implant components; fixtures, abutments and screws is fatigue failure. This occurs as a result of cyclic functional loading; the magnitude of which may be well below the ultimate strengths of the components. Good clinical practice and adherence to sound biomechanical principles of prosthesis design should minimise the risks of fatigue failure, although component design may play a role.

Catastrophic failure can be modelled in-vitro with some confidence by the application of a single load cycle applied by a calibrated testing system until failure occurs. Fatigue loading is much more complex to model in-vivo and it can be difficult to extrapolate to clinical behaviour from such findings. Clinical study of failure specimens may be helpful in identifying possible contributory factors. An international standard (ISO 14801:2003(E)) exists for fatigue testing of endosseous dental implants. It is most useful for comparing implants of different designs or sizes.

The published standard carries an important caveat not included in the similar standard for the testing of orthopaedic prostheses: ‘Whilst it simulates the functional loading of an endosseous dental implant body and its premanufactured prosthetic components under ‘worst case’ conditions, the standard is not applicable for predicting the in-vivo performance of an endosseous dental implant or prosthesis.’

A number of workers have identified the potential risks of fatigue failure, particularly in single tooth restorations (Marinello et al. 1997; Schmitt and Zarb 1993). Henry et al. (1996), however, reported a very low incidence of fatigue fracture in a prospective five year study. Accurate seating of the abutment on an implant is critical (Binon, 2000a). A loss of preload in the abutment screw can result in loosening with wear and fracture (Binon 2000b)(Haak et al. 1995) in the flat-on-flat systems.

If failure of an implant component occurs as a result of fatigue it is essential to understand the mode of failure. The loss of preload in an abutment screw (possibly due to an incorrect torque at placement) may result in loosening of the implant abutment connection and the subsequent development of unexpected bending loads causing fracture of the implant. Clear interpretation of the steps leading to failure by making a thorough examination of fractured components is the only way to correct diagnosis. It therefore becomes clear that simulation and modelling in-vitro of multifactorial dynamic fracture modes is a great challenge.
A number of workers have developed sophisticated and elegant loading machines designed specifically to model masticatory function (Sakaguchi et al. 1986; Bates, Stafford and Harrison, 1975) for the study of wear and less frequently fracture. However, a simple uniaxial loading regimen can be useful in comparing implant performance (ISO 14801:2003(E)).

This investigation was prompted by the only reported clinical failure of a specific implant (Neoss system; Neoss Limited, Harrogate, England). The fracture of the implant flange occurred and was identified some 3.5 years after placement. The implant was one of three comprising a bar retained overdenture with electroformed denture. Figure 1 illustrates the fractured implant in-situ. The fractured element was removed and examined under scanning electron microscopy (SEM; Figure 2).

As a result of these investigations it was deduced that the most likely aetiology of failure was fatigue fracture. In order to better understand the events contributing to this single failure an in-vitro investigation was carried out according to the international standard test method ISO14801. The National Swedish Test House (Sveriges Tekniska Forskningsinstitut) was commissioned to undertake this study.

**METHOD**

Twenty four implants (Neoss Limited; Harrogate UK) in two groups of diameter 3.5mm and 4.0mm were embedded in a specimen holder with a heat curing resin (Epotek 353ND, CA, Modulus 4.0GN/m²) and at a depth in accordance with ISO14801. Abutments (identification number 10547) were attached to each implant with a gold abutment screw (identification number E8140) and torqued to 32Ncm⁻¹ using a calibrated torque wrench. A hemispherical loading member was attached to each abutment. The fatigue tests were performed in a servohydraulic test machine (Instron 1341; Instron Ltd, Uxbridge, England) as constant load amplitude tests with $R=0.1$ at 15 Hz. $R$ is defined as $F_{\text{min}}$ divided by $F_{\text{max}}$. Fracture was defined as a visible...
unrecovered deformation, i.e. when a deflection of approximately five millimetres of the loading point occurred. The initial position of the deflection is measured at Fmax. The test set-up is shown in figure 3.

The aluminium cylinder of the test specimen was mounted in a holder of stainless steel so that the force was applied at 30° to the centre axis of the specimen top. The force was applied to the polished loading point of the specimen top using a flat ended cylindrical rod with a concave recess mounted onto the load cell. The holder was mounted on the base plate of a cylindrical stainless steel test reservoir and positioned to ensure vertical load application. The horizontal position of the specimen holder was adjusted at the start of the test to ensure the load application was in a vertical line beneath the load cell. The loading parameters are listed in Tables 1 and 2. Load cycling was carried out on each specimen for 5 million cycles (‘run out’) or until fracture occurred. Fracture was classified as A – of the abutment screw, B – of the abutment screw and crack on the fixture or C- fracture on the fixture and epoxy (Epotek 353ND; Epotek Corp. Ca).

RESULTS

The results from the fatigue tests are tabulated in Table 1 for 3.5mm implants and Table 2 for 4.0mm implants. Results from fatigue tests are plotted as a Wohler graph (Fmax vs. Numbers of cycles in logarithmic scale) for the 3.5mm implants (Figure 4) and 4.0mm implants (Figure 5). In the Wohler graph the formula $N = A \times F_{max}^B$ describes the relation between Fmax and the numbers of cycles to fracture. The constants A and B are calculated as a least square fit in a log-log plot where $F_{max}$ is the controlled variable and N (the numbers of cycles to fracture) is the obtained variable. The results are tabulated in Table 3.
Figure 4. Wohler graph for the fatigue test on fixture 3.5 mm diameter

![Wohler graph for fixture 3.5 mm diameter](image)

Table 3. Linear regression parameters for fractured specimens

<table>
<thead>
<tr>
<th>Fixture</th>
<th>Constant A</th>
<th>Constant B</th>
<th>R²</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.5 mm dia.</td>
<td>1.433 x 1043</td>
<td>-14.72</td>
<td>-0.75924</td>
</tr>
<tr>
<td>4.0 mm dia.</td>
<td>1.524 x 1050</td>
<td>-17.25</td>
<td>-0.76404</td>
</tr>
</tbody>
</table>

Figure 5. Wohler graph for the fatigue test on fixture 4.0 mm diameter

![Wohler graph for fixture 4.0 mm diameter](image)
DISCUSSION

Fatigue testing is the repeated application of cyclic mechanical loading to a specimen with the intention of simulating accelerated function leading to possible fracture. The complex loading conditions and variables in occlusal function make accurate simulation of such loading extremely difficult (De Boever, 1978; Kraus, Jordan and Abrams, 1969; Haraldson, Carlsson and Ingervall, 1979). Indeed the ISO standard (14801:2003(E)) states that whilst the standard is designed to simulate the functional loading of an endosseous dental implant body and its premanufactured prosthetic components under worst case conditions the standard is not applicable for predicting the in-vivo performance of an endosseous implant. The purpose and value of such testing may therefore be open to question. However such testing is useful in giving indicators of possible weaknesses in design and production and for comparison between products and implant systems (Strub & Gerds, 2003).

The most useful parameter in the results is the lowest value of $F_{max}$ at which an implant abutment complex ceases to fail and performs a repeatable runout between specimens. In this study the value was 280N for 3.5mm diameter implants Published data is relatively scarce, however Straumann (2007) have published data comparable data for 3.5mm implants obtained by the ISO 14801 standard (Figure 6). It is also important to understand the mode and pattern of failure that occurs. Clinically, for example, abutment screw fracture may have less serious implications in screw retained system than in cemented restorations (Drago 1995). Fracture of an implant will nearly always have serious consequences. Clinical examination of the case described previously with the one and only fractured Neoss illustrates a number of useful points.

Figure 7 illustrates the over denture which was being supported by the fractured implant. There were originally four implants placed, two each in the premolar regions. One implant was lost shortly after prosthetic reconstruction and not replaced. Three and a half years later the remaining implant was the one that fractured on the left side. This illustrates significant mesial and distal cantilevers and the clinician described the aetiology as biomechanical overload. Furthermore examination of the fractured implant margin (Figure 8.) indicates burnishing of the top face (Cibirka et al. 2001). This would only occur if the abutment screw had loosened allowing unnoticed movement of the implant and abutment at the interface resulting in high loads being applied to the implant margin leading to the failure (Gratton, Aquilino and Stanford, 2001; Khraisat et al. 2004).

In conclusion fatigue testing of dental implants and components alone is of limited value. In combination with clinical data and an understanding of the aetiology of failure useful data can be obtained to predict implant performance. The fatigue data for 3.5 and 4.0mm diameter Neoss implant compare favourably with other industry standard implant systems of unspecified size or function.
Figure 8. SEM fractured implant face (a) of burnished implant/abutment interface (b)

ACKNOWLEDGEMENTS

The authors would like to thank Klas Johansson, Technical Manager and Jukka Holappa, Technical Officer of SP Sveriges Tekniska Forskningsinstitut, Building Technology and Mechanics – Solid Mechanics and Structures for performing the fatigue testing.

REFERENCES


INTERNATIONAL STANDARD; Dentistry — Fatigue test for endosseous dental implants ISO 14801:2003(E)


Straumann. TARGET 2007/4 : 24-25
