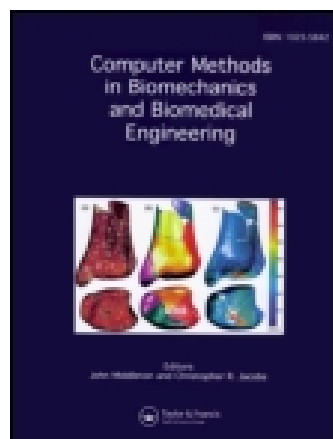


This article was downloaded by: [Tulane University]

On: 29 September 2014, At: 06:35

Publisher: Taylor & Francis

Informa Ltd Registered in England and Wales Registered Number: 1072954 Registered office: Mortimer House, 37-41 Mortimer Street, London W1T 3JH, UK



## Computer Methods in Biomechanics and Biomedical Engineering

Publication details, including instructions for authors and subscription information:

<http://www.tandfonline.com/loi/gcmb20>

### Application of the finite element method in dental implant research

R. C. Van Staden<sup>a</sup>, H. Guan<sup>a</sup> & Y. C. Loo<sup>a</sup>

<sup>a</sup> School of Engineering, Griffith University Gold Coast Campus, PMB 50 Gold Coast Mail Centre, Queensland, 9726, Australia

Published online: 07 Dec 2006.

To cite this article: R. C. Van Staden, H. Guan & Y. C. Loo (2006) Application of the finite element method in dental implant research, *Computer Methods in Biomechanics and Biomedical Engineering*, 9:4, 257-270, DOI: [10.1080/10255840600837074](https://doi.org/10.1080/10255840600837074)

To link to this article: <http://dx.doi.org/10.1080/10255840600837074>

PLEASE SCROLL DOWN FOR ARTICLE

Taylor & Francis makes every effort to ensure the accuracy of all the information (the "Content") contained in the publications on our platform. However, Taylor & Francis, our agents, and our licensors make no representations or warranties whatsoever as to the accuracy, completeness, or suitability for any purpose of the Content. Any opinions and views expressed in this publication are the opinions and views of the authors, and are not the views of or endorsed by Taylor & Francis. The accuracy of the Content should not be relied upon and should be independently verified with primary sources of information. Taylor and Francis shall not be liable for any losses, actions, claims, proceedings, demands, costs, expenses, damages, and other liabilities whatsoever or howsoever caused arising directly or indirectly in connection with, in relation to or arising out of the use of the Content.

This article may be used for research, teaching, and private study purposes. Any substantial or systematic reproduction, redistribution, reselling, loan, sub-licensing, systematic supply, or distribution in any form to anyone is expressly forbidden. Terms & Conditions of access and use can be found at <http://www.tandfonline.com/page/terms-and-conditions>

# Application of the finite element method in dental implant research

R. C. VAN STADEN, H. GUAN\* and Y. C. LOO

School of Engineering, Griffith University Gold Coast Campus, PMB 50 Gold Coast Mail Centre, Queensland 9726, Australia

(Received 5 December 2005; revised 27 May 2006; in final form 27 June 2006)

This article provides a review of the achievements and advancements in dental technology brought about by computer-aided design and the all powerful finite element method (FEM) of analysis. The scope of the review covers dental implants, jawbone surrounding the implant and the biomechanical implant and jawbone interaction. Prevailing assumptions made in the published finite element analysis (FEA) and their limitations are discussed in some detail which helps identify the gaps in research as well as future research direction.

**Keywords:** Dental implant; Implant–jawbone interaction; Design; Finite element analysis

**Subject Classification:** Engineering application in Dental Implant Research

## 1. Introduction

Development of an ideal substitute for missing teeth has been one of the long-term aims of dentistry. A dental implant is a biocompatible screw-like titanium “fixture” that is surgically placed into the jawbone. Figure 1 provides some detail of a typical implant design† and figure 2 shows its orientation within the jawbone‡. The implant is anchored in the jawbone. An implant post or abutment and permanent tooth can be attached in a variety of designs.

The long-term benefits of dental implants include improved appearance, comfort, speech and self-esteem. With the dental implant, the patient can eat more conveniently and the inconvenience of and at times embarrassment caused by removable partial and full dentures can be eliminated. In addition, the implant is able to protect the remaining natural teeth, stop bone loss and restore facial skeletal structure. As far as the costs are concerned, implants have shown to be less expensive overall than other types of prostheses such as crown and bridge restorations. Currently, over three hundred thousand implants are in use in the United States alone (O’Brien 1989).

Worldwide statistics show a high success rate of implantation in excess of 95% if the implants are correctly

designed, manufactured and inserted. Implants are expected to be functioning for a life-long period. This is justified by the fact that the survival rate at 15 years is as high as 90% if proper and professional cares are taken. Despite all these advantages, only 15% of Australians have been supplied with single or multiple dental implants. This is mainly due to the lack of general understanding by the dentists of the stress characteristics of implant and jawbone and the biomechanical behaviour of their interaction. Another reason is the lack of clinical skills in consequence of complicated and not well-known insertion technique.

The success of a dental implant depends on a variety of factors including the design of the abutment and technique by which the abutment screw is placed into the implant. Major causes of implant failure are due to insufficient biomechanical bonding between the implant and the surrounding jawbone and also implant tooth fixtures or abutment failure. Insufficient biomechanical bonding might be initiated by, firstly, insufficient osseointegration because of the jawbone not accepting the implant. In this case, the only alternative, for a single missing tooth, would be to use a bridge. Secondly, after insertion, the implant might fail due to the lack of hygiene or overload. A solution would be to insert a new implant after the jawbone has restored itself. Other implant failures could

\*Corresponding author. Email: h.guan@griffith.edu.au

†Neoss Australia Pty Ltd.

‡University of Southern California. Available online at: [www-hsc.usc.edu](http://www-hsc.usc.edu) (accessed 25 November 2004).

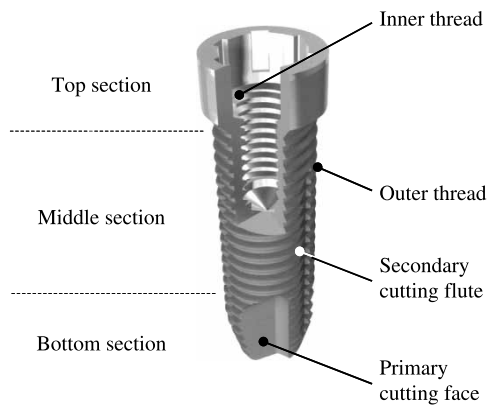


Figure 1. Three-dimensional view of dental implant produced by Neoss (Courtesy of Neoss Australia Pty Ltd).

be due to inaccurate design of the tooth fixture leading to overload of the abutment screw and eventually micro fractures within the implant.

Significant similarities exist between dental implants and some mechanical components, for example the bolts used in vehicle engines. The success of the bolt within the engine relies on similar factors to that of the dental implant. These factors might include load applied to bolt/implant and the stress profile within the engine block/jawbone. The dental implant however differs from the engine bolt in that the jawbone is not a rigid body like the engine block. The jawbone usually exhibits an unpredictable biomechanical reaction to the dental implant which complicates the entire implantation and integration process.

The phenomenon of bone response to a foreign insert, such as an implant, has not been studied adequately and the long-term effects of such stresses are still unclear. A thorough understanding of this phenomenon might lead to a reduction in the undesirable stresses produced within the jawbone. Bone is a self-adaptive material, which means that when the surrounding stress is changed, the bone tissue structure is adjusted by itself to suit the new loading environment. This is known as the bone remodelling. Such remodelling includes both internal modification of apparent bone density and external alteration of bone shape. Bone remodelling induced by the change of normal biological stress is one of the most important factors causing implant failure. This is the so-called stress shielding effect. Using the finite element method (FEM) and parameterised optimum design technique, the stress shielding effect can be minimised to a maximum extent by performing multi-parameter optimisation of implant, thereby guaranteeing the success rate of implantation.

The FEM is a numerical method of analysis for stresses and deformations in structures of any given geometry. The structure is discretized into the so called "finite elements" connected through nodes. The type, arrangement and total number of elements affect the accuracy of the results. The FEM has become one of the most successful engineering computational methods and most useful analysis tool since the 1960s (Ergatoudis *et al.* 1968,

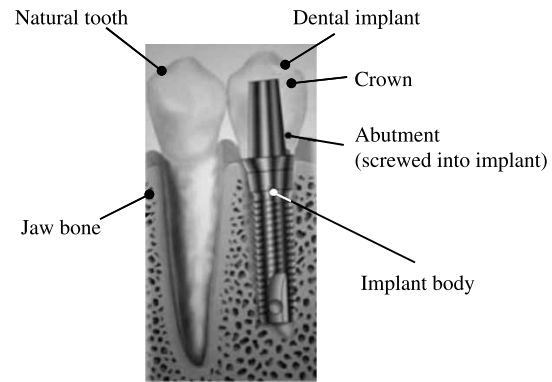


Figure 2. Cross-sectional view of a natural tooth and a dental implant (www-hsc.usc.edu).

Rzemieniecki 1969). It is showing overwhelming capability and versatility in its application in dentistry (Ismail *et al.* 1987, Rieger *et al.* 1990, Waskewicz *et al.* 1994, Canay *et al.* 1996, Stegaroiu *et al.* 1998, Akpinar *et al.* 2000, DeTolla *et al.* 2000, Akca and Iplikcioglu 2001, Geng *et al.* 2001, Akca *et al.* 2002, Nagasao *et al.* 2002, 2003, Pierrisnard *et al.* 2002, Plikciolu and Akça 2002, Ishigaki *et al.* 2003, Watanabe *et al.* 2003, Eskitascioglu *et al.* 2004, Himmlova *et al.* 2004). This paper reviews past and current practices in the finite element analysis (FEA) of dental implants. The achievements and limitations of the existing analysis are discussed and the gap in research is identified. Future research directions are also recommended with particular emphasis on the stress evaluation and design optimisation associated with the implants.

## 2. Finite element modelling and analysis techniques

The jawbone and implants are very complicated structures. It is difficult to establish an accurate and valid three-dimensional (3D) finite element model using conventional modelling techniques. Two-dimensional (2D) representations of implants and jawbone structures were often assumed in previous studies. Some of which also failed to recognise the difference between the cortical and trabecular bones. As such, the calculated results are often far from the actual situation for 2D analysis, hence they cannot be used as a useful guidance to the implant treatment.

Generally, previous analysis were conducted by firstly constructing a finite element model, followed by specifying appropriate material properties, loading and boundary conditions so that the desired scenarios can be accurately simulated. Various engineering software packages have been employed to model and simulate both the implant and jawbone.

Selection of either 2D or 3D analysis would depend on the software capabilities and also the required accuracy of the results. Tables 1 and 2 show the software specification, analysis level as well as the finite element meshing

Table 1. Summary of analysis software and analysis level.

Source	Papavasiliou <i>et al.</i> (1997)	Pierrisnard <i>et al.</i> (2002)	Patra <i>et al.</i> (1998)	Canay <i>et al.</i> (1996)	Meijer <i>et al.</i> (1993)	Borchers and Reichart (1983)	Zarone <i>et al.</i> (2003)	Lewinstein <i>et al.</i> (1995)	Menicucci <i>et al.</i> (2002)
Software	COSMOS/M Ver 1.61, (SRAC)	CADSAP	SDRC I-DEAS, Hypermesh, Ansys	SAP 90	CAEDS	SAP IV	AutoCAD, SDRC I- DEAS VI	ADINA	ADINA
Analysis level	3D	3D	2D	2D	3D	3D	3D	2D	2D and 3D

schemes used by various researchers during their analysis of dental implants.

Accurate and efficient modelling can provide insight and understanding of the complicated nature of a dental implant that is surrounded by the jawbone. The success of modelling depends on the accuracy in simulating the geometry and surface structure of the implant, the material characteristics of the implant and jawbone, the loading and support conditions as well as the biomechanical implant–jawbone interface.

### 3. Assumptions made in existing finite element analysis

FEM has become one of the most successful analysis methods used for solving dental related problems. Not unlike many other finite element applications, certain assumptions need to be made in dealing with complicated implant, jawbone and implant–jawbone interaction problems. In general, five major simplifications were made in existing analysis of implant and jawbone structures:

- (1) The 2D representation of implant and jawbone structures is adopted based on either the assumption of axial symmetry of loads or geometry (Lewinstein *et al.* 1995, Canay *et al.* 1996, Patra *et al.* 1998). A majority of the 2D studies claim that the translation of the clinical condition to a 2D model gives sufficient insight into the behaviour of the jawbone around implants. However, the stress in the jawbone predicted by a 2D model is less accurate than that predicted by its 3D counterpart (DeTolla *et al.* 2000).

- (2) In most research reported to date, axially applied static loads have been assumed instead of the more realistic dynamic–cyclic loads directed at the occlusal angle encountered in the jawbone during mastication of food (Geng *et al.* 2001).
- (3) The fact that the biomechanical reaction of the jawbone differs for each patient means that it is impossible to model the percentage of osseointegration accurately. A fixed bond between the jawbone and implant along the entire interface has been assumed. This implies that under any loading, the relative motion between jawbone and implant does not occur (Papavasiliou *et al.* 1997, Kohal *et al.* 2002, Menicucci *et al.* 2002).
- (4) An assumption of homogeneous, linear, elastic material behaviour for the jawbone is used which is characterised by a single Young's modulus and Poisson's ratio (Mihalko *et al.* 1992, Nishihara and Nakagiri 1994, Lewinstein *et al.* 1995).
- (5) The choice of implant design for each patient depends solely on the identification, by the clinician, of the patient's bone type. An incorrect choice might lead to an implant failure. Only four different jawbone structures have been evaluated in the previous research. This might also have led to unrealistic results (Rho *et al.* 1993).

### 4. Modelling of implant

Modelling of the implant involves accurate representation of the implant geometry, taperage, material properties, loading conditions and implant surface structure. Following this, a redesign of the implant geometry can be performed to achieve an optimum stress profile in the

Table 2. Summary of finite element meshing schemes.

Source		Papavasiliou <i>et al.</i> (1997)	Pierrisnard <i>et al.</i> (2002)	Canay <i>et al.</i> (1996)	Borchers and Reichart (1983)	Zarone <i>et al.</i> (2003)	Lewinstein <i>et al.</i> (1995)	Ishigaki <i>et al.</i> (2003)
Element type		Brick (not symmetrical)	Brick	Brick	Brick (hexahedral)	Brick (solid linear and parabolic wedge)	Brick	Brick
Implant and jawbone	Element number	480	4322–7766 (6–12 mm in length)	311	1129	2356	6441	1731
	Node number	*N/A	*N/A	295	1473	*N/A	26351	2269

\*N/A, information not available.

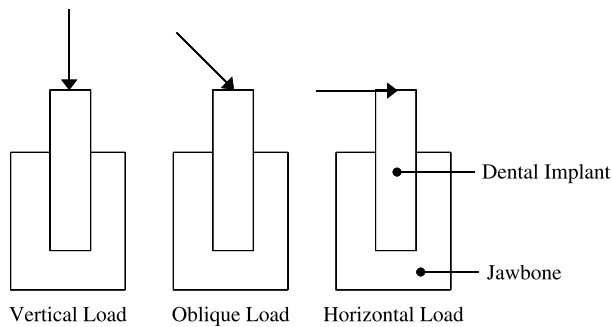


Figure 3. Different loading directions applied to a simplified implant and jawbone model.

surrounding jawbone. Creating an accurate analytical model of a dental implant, using appropriate engineering software, is essential in producing realistic and reliable solutions.

Modelling assumptions and software limitations might have led to a number of inaccuracies within the obtained results. Figure 3 shows a simplified dental implant model, together with the surrounding jawbone, with applied loads in three directions. It is often assumed that the dental implant is rectangular in shape and modelled in 2D, with no cutting edges, screw thread and surface structure. A fixed bond (100% osseointegration) is also often assumed between the implant and jawbone. It was found common in many publications to assume either a vertical, oblique, occlusal, horizontal or a combination of the loads applied to the implant (Borchers and Reichart 1983, Meijer *et al.* 1993, Canay *et al.* 1996, Papavasiliou *et al.* 1997, Patra *et al.* 1998, Richter 1998, Akpınar *et al.* 2000, O'Mahony *et al.* 2000, Geng *et al.* 2001, Menicucci *et al.* 2002, Pierrisnard *et al.* 2003, Tada *et al.* 2003, Yokoyama *et al.* 2004).

#### 4.1 Implant geometry

The implant geometry is an important criterion to evaluate during any analysis that considers the biomechanical influence that the implant has on the surrounding jawbone.

Siegele and Soltesz (1989) investigated the stress distribution generated within the jawbone surrounding different types of dental implants (cylindrical, conical, stepped, screw-shaped, hollow cylindrical) by means of the FEM. The results demonstrate that different implant shapes lead to significant variations in stress distributions in the jawbone. In particular, implant surfaces with very small radii of curvature (conical) or geometric discontinuities (stepped) imply distinctly higher stresses than smoother shapes (cylindrical, screw-shaped). On the other hand, Holmgren *et al.* (1998) reported that a stepped cylindrical design is most desirable for stress distribution in the surrounding jawbone.

The implant length and diameter has a significant influence on the stress distribution, within the surrounding jawbone (Aparicio and Orozco 1998, Ivanoff *et al.* 1999, Misch 1999, Zhang *et al.* 2000, Iplikcioglu and Akca

2002, Plikçiolu and Akça 2002, Tawil *et al.* 2002, Pierrisnard *et al.* 2003, Yokoyama *et al.* 2004). Finite element simulation of stress distribution around implants can be used to determine the optimum length and diameter of the implants that would best dissipate stresses induced by the implantation. Himmlova *et al.* (2004) used FEA to compute values of von Mises stress at the implant–bone interface for all variations in length and diameter of implants. Maximum stress areas were identified to be located around the implant neck. The maximum decrease in stress (31.5%) was found for implants with a diameter ranging from 3.6 to 4.2 mm. Further stress reduction for the 5.0-mm implant was only 16.4%. An increase in the implant length also led to a decrease in the maximum von Mises equivalent stress values; the implant length, however, was not as influential as that of implant diameter. Note that the length of implant ranges from 8 to 18 mm (Himmlova *et al.* 2004).

Short implants present superior failure rates. Pierrisnard *et al.* (2003) investigated the influence of implant length and bicortical anchorage on implant stress distribution. The aim of this theoretical study was to assess to what extent the implant length and bicortical anchorage affect the way stress is transferred to the implant components, the implant abutment and the surrounding bone. Stress analysis was performed using the FEM and a 3D linear elastic model was generated. All implants modelled were of the same diameter (3.75 mm) but varied in length, at 6–12 mm. Each implant was modelled with a titanium abutment screw and abutment, a gold cylinder and prosthetic screw and a ceramic crown. The implants were seated in a supporting bone structure consisting of cortical and cancellous bone. It was concluded that the maximum bone stress was virtually constant, independent of implant length and bicortical anchorage. The maximum implant stress, however, increased somewhat with implant length and bicortical anchorage.

Wide-bodied implants placed in the posterior jaw can suffer a significantly elevated risk of implant failure compared to “regular-diameter” implants. This may relate to either implant design or the relative relationship of implant to surrounding bone dimensions. Shin *et al.* (2004) conducted a retrospective study documenting a 5-year cumulative survival rate (CSR) of 5-mm-diameter wide-bodied implants in posterior jaws as related to identified risk factors and relative host bone site dimensions. During this study, it was found that wide-bodied implants had a CSR of 80.9%, while “regular-diameter” implants had a CSR of 96.8%. Differences between wide-bodied and “regular-diameter” implants was found to be statistically significant. Multivariate analysis demonstrated a significant predictive relationship between overall CSR and the ratio of implant to bone volume. A significant difference found between the CSR of the “regular-diameter” and wide-bodied implants might be due to the differences in the damping capacity of the initial interfacial tissue. The difference between the CSR found in the maxilla (Mx) and the mandible (Md) has not



Table 3. CSR of wide-bodied implants (Shin *et al.* 2004).

Source	CSR (%)
Aparicio and Orozco (1998)	Mx: 97.2 Md: 83.4
Ivanoff <i>et al.</i> (1999)	Mx: 86.3 Md: 73.0
Tawil <i>et al.</i> (2002)	P: 96.9
Shin <i>et al.</i> (2004)	P: 80.9

Note: Mx, maxilla; Md, mandible; and P, posterior Mx and Md.

been discussed although it can be observed that generally CSR in Mx is larger than that in Md. The CSR values of wide-bodied implants which was found in various other studies, are shown in table 3.

An increase in the diameter of the implant led to a decrease in the CSR value, indicating that the optimum diameter of an implant should be less than that of the wide-bodied implants used in Shin *et al.*'s work. This suggests that relative determinants of critical bone volume to implant dimensions may need to be considered when planning implant surgery. Implant characteristics found in various studies are shown in table 4.

Many researchers recognized the influence that the implant geometry and the quality of the surrounding jawbone have on the overall success of the implantation, Ivanoff *et al.* (1999) found that a relationship could be seen between implant failure and implant diameter for 4 mm diameter implants evaluated between 3 and 5 years following insertion. However, no relationship could be seen between implant failure and jaw type, or bone quality and quantity for implant diameters of 3.75 and 5 mm between 3 and 5 years following insertion and also for 4 and 5 mm diameter implants after 5 years following insertion. Neither was any relationship seen between marginal bone loss and bone quality and quantity, implant diameter, or jaw type for implant diameters of 3.75 and 5 mm between 3 and 5 years following insertion and also for 4–5 mm diameter implants after 5 years following insertion, when tested by multiple linear regression analysis.

## 4.2 Taperage

Dental implants might have a degree of taperage applied to either the entire implant or just the bottom section of the implant. Tapered implants enhances primary stability by compressing the surrounding jawbone during and after implantation. The tapered body also facilitates implant placement in convergent roots, in areas with labial concavities and in immediate extraction sites.

Mailath *et al.* (1989) compared cylindrical and conical implant shapes exposed to physiological stresses and examined the occurrence of stress concentrations at the site of implant entry into the jawbone. It was reported that cylindrical implants produced a more desirable stress profile than the conical shaped counterparts. The need for tapered implants have been identified due to the fact that the survival rate of oral implants in soft-quality bone were demonstrated to be inferior to that of implants inserted in good-quality bone.

Astrand *et al.* (2003) showed that a possible way to increase the survival rate in soft-quality bone may be to use a tapered implant. Such an implant has been developed and manufactured by Nobel Biocare AB, Gothenburg, Sweden (<http://www.nobelbiocare.com>) (Nobel Biocare). The aim of their study was to compare the outcome of using the tapered Branemark System Mark IV fixture with that of using earlier Branemark fixtures in a controlled prospective study. It was concluded that with regard to survival rate and marginal bone level changes, no differences could be demonstrated between the Mark IV tapered implant and the Branemark implants used earlier. However, compared with earlier results of Branemark implants in soft-quality bone, the Mark IV (increased taperage) implant demonstrated an improved survival rate. Friberg *et al.* (2003) investigated through practical experimentation the tapered implants as compared to the straight cylindrical ones. The tapered implant design resulted in an increased primary stability which may be important when placing implants in jaw regions of type 4 bone.

Table 4. Dental implant model specification.

Source		Papavasiliou <i>et al.</i> (1997)	Pierrisnard <i>et al.</i> (2002)	Patra <i>et al.</i> (1998)	Canay <i>et al.</i> (1996)	Meijer <i>et al.</i> (1993)	Borchers and Reichart (1983)	Zarone <i>et al.</i> (2003)	Lewinstein <i>et al.</i> (1995)
Implant design		IMZ (Interpore International, Irvine, CA, USA)	Nobel Biocare AB (Branemark System)	BUD (parallel profile thread), Branemark (tapered thread)	ITI Bonefit (Straumann)	*N/A	Anchor-type endosteal	*N/A	Branemark System (IL system)
Implant size	Length (mm)	11	6–12	13	*N/A	20	*N/A	13	6
	Diameter (mm)	4	3.7–5	*N/A	*N/A	3.7–5	*N/A	3.2–5	3.7–5
Implant material		Titanium, Ti (pure)	Titanium, Ti (pure)	Titanium, Ti (pure)	Titanium, Ti (pure)	*N/A	Al <sub>2</sub> O <sub>3</sub> –ceramic	*N/A	*N/A

\*N/A, information not available.

O'Sullivan *et al.* (2004) conducted a study to analyse the mechanical performance and the primary and secondary stability characteristics of endosseous titanium implants with 1 degree (EXP1) and 2 degrees (EXP2) of taper when compared with the standard Branemark design (<http://www.nobelbiocare.com>) (Nobel Biocare). At placement, significantly higher insertion torque (IT) was needed to insert the EXP implants. Resonance frequency analysis (RFA) has been used to quantitatively measure implant stability over time. It has long been a recognized need among dental professionals involved in dental implant therapy (Meredith *et al.* 1997). The measurement is taken with a transducer screwed onto the implant, the piezo elements of which are caused to oscillate. The device records the resonance frequency produced from the implant–bone interface which is displayed graphically. The oscillation of the implant transducer element is recorded as the implant stability quotient (ISQ) (Osstel). Successful implantations indicate an increase in the recorded RFA value, between the period of surgical insertion and the connection of the abutment. A decrease in the RFA recorded value causes a marginal bone loss, which in turn would yield unsuccessful implantation. RFA values were significantly higher for EXP1 implants placed in the tibia but not in the femur. In pooling data from the femur and tibia, there was a significant difference. The EXP2 implants failed to insert fully and demonstrated a lower RFA value than may have been expected due to the exposed threads, although this difference was not statistically significant. The results obtained by O'Sullivan *et al.* (2004) showed that the EXP1 resulted in a better primary stability compared with the standard Branemark design. There was no evidence that the tapered design caused negative bone tissue reactions. Stability due to the taperage was gained during the healing period.

#### 4.3 Material properties of implant

The biomechanical properties of an implant are very different to the original tissue. The design and manufacture of artificial materials that are aimed at being used within the human body, is becoming increasingly challenging.

Titanium and bio-ceramic materials, such as hydroxyapatite (HA) are extensively used as fabrication materials for dental implant due to their high compatibility with hard tissue and living bone (Hedia and Mahmoud 2004). Titanium has reasonable stiffness and strength while HA has low stiffness, low strength and high ability to reach full integration with living bone. A number of material properties have been adopted for titanium in previous

analysis. These are presented in table 5, where the Young's modulus of titanium has been taken from 102.2 to 113.8 GPa, and the Poisson's ratio, between 0.29 and 0.35, based on different material testing.

In order to achieve adequate dental implantation of the biomaterial, full integration of the implant with living bone must be guaranteed. Uniform stresses of an optimum magnitude in the implant and jawbone must be satisfied to increase the life of the implant and prevent bone resorption. Hedia and Mahmoud (2004) investigated functionally graded material (FGM) to achieve the above advantages. The optimal materials of the FGM dental implant are found to be a combination of HA and titanium. The finite element investigations have shown that the maximum stress in the bone for the HA/titanium FGM implant has been reduced by about 22 and 28%, respectively compared to currently employed titanium and stainless steel dental implants.

Moroi *et al.* (1993) found that the HA coating reduced the heat conduction to the surrounding tissue because of its low thermal conductivity. However, thermal stress was induced by thermal expansion of the HA and titanium core. This might influence the success of HA-coated implants because the biomechanical properties of ceramics are sufficiently poor leading to insufficient tensile and shear stresses in the surrounding jawbone. The biomechanical integration between living bone and ceramic are not as successful as that of pure titanium. In addition, this tendency becomes more pronounced when the HA-coating surface extends beyond the cortical bone. It is important to fully understand the mechanical behaviour of a natural tooth, in order to improve dental implant performance within the surrounding jawbone. Table 6 shows the material properties of a natural tooth, adopted by various researchers.

The Young's modulus of enamel summarised in table 6 has an unexplained variance that could possibly be due to the fact that different testing techniques were employed. For example, by altering the angle of inclination, with respect to the vertical direction, at which the impact test is used will drastically affect the obtained results for the strength of the enamel. The strength of the enamel will be significantly less if it was impact tested parallel to the tubules of the enamel. For other parts of a tooth, Menicucci *et al.* (2002) found that Mucosa had a Young's modulus,  $E$  of 1 MPa and a Poisson's ratio,  $\nu$  of 0.37. It was also found that Pulp had a Young's modulus,  $E$  of 2 MPa and a Poisson's ratio,  $\nu$  of 0.45. The natural tooth is the ultimate design for its operation and to truly understand its material properties and function is the key

Table 5. Assumed material properties for titanium used in implant modelling.

Source	Papavasiliou <i>et al.</i> (1997)	Pierrisnard <i>et al.</i> (2002)	Patra <i>et al.</i> (1998)	Canay <i>et al.</i> (1996)	Meijer <i>et al.</i> (1993)	Zarone <i>et al.</i> (2003)	Lewinstein <i>et al.</i> (1995)	Menicucci <i>et al.</i> (2002)
Young's modulus, $E$ (GPa)	102.2	110	110	113.8	103.4	110	110	103.4
Poisson's ratio, $\nu$	0.35	0.3	0.33	0.35	0.35	0.29	0.33	0.35

Table 6. Material properties of natural tooth.

Source		Menicucci <i>et al.</i> (2002)	Davy <i>et al.</i> (1981)	Wright and Yettram (1979)	Farah <i>et al.</i> (1989)	Farah <i>et al.</i> (1974)	Reinhardt <i>et al.</i> (1984)	MacGregor <i>et al.</i> (1980)
Enamel	Young's modulus, $E$ (GPa)	84100	41400	46900	82500	84000	*N/A	*N/A
	Poisson's ratio, $\nu$	0.2	0.3	0.3	0.33	0.33	*N/A	*N/A
Dentin	Young's modulus, $E$ (GPa)	18600	*N/A	*N/A	*N/A	*N/A	18600	18000
	Poisson's ratio, $\nu$	0.31	*N/A	*N/A	*N/A	*N/A	0.31	0.31

\*N/A, information not available.

to future implant enhancement. The material property information provided in tables 5 and 6 should be used as a guide in creating the finite element models of dental implant and jawbone.

#### 4.4 Loading

Masticatory forces acting on dental implants can result in undesirable stress within the surrounding jawbone which in turn can cause bone rejection to the implant, leading to eventual failure of the implant. In order to produce accurate predictions of the implant–jawbone behaviour, it is essential to determine realistic loading magnitudes and directions. Richter (1998) investigated the actual loads experienced by teeth during chewing. Four main types of food (rubbery confection, pieces of sausage, carrots and crackers) were examined for their effects on the load experienced by teeth during chewing. Table 7 shows the typical load level experienced in the vertical, transverse and mesiodistal directions, as indicated in figure 4.

Haraldson *et al.* (1988) evaluated the mastication forces and chewing efficiency in nine patients, whom were treated with overdentures on osseointegrated implants in the Md. The bite force was measured during gentle biting, biting as when chewing and biting with maximal effort. All subjects improved subjectively as well as clinically after treatment. The bite force during gentle biting increased on average from 17.3 N before treatment to 24.0 N one year after treatment. A corresponding improvement of biting as when chewing was also found; from on average 24.0 N before to 38.7 N after treatment. The maximal bite force increased from on average 74.6 N at the baseline examination to 131.5 N at the one year follow-up. The chewing efficiency improved from the

confidence interval,  $C_i = 4$  (median value) before treatment to  $C_i = 2.8$  (median value) after treatment. Which means that the middle value of the patients who experienced an improvement in chewing efficiency improved by a factor of 1.2. It is concluded that treatment with an overdenture supported by osseointegrated implants in the Md improves oral function compared to the situation before treatment.

To successfully replicate the clinical situation, that an implant might encounter, it is important to understand the loading that is induced by the jawbone muscles unto the implant. Koolstra *et al.* (1988) constructed a 3D mathematical model of the human masticatory system, containing 16 muscle forces and two joint reaction forces. Maximum possible bite forces were computed using optimisation techniques; the optimisation criterion used was to minimize the relative activity of the most active muscle. The model predicted that at each specific bite point, bite forces can be generated in a wide range of directions and that the magnitude of the maximum bite force depends on its direction. The relationship between bite force direction and its maximum magnitude depends on bite point location and mandibular position. In general, the direction of the largest possible bite force does not coincide with the direction perpendicular to the occlusal plane, as illustrated in figure 5.

A significant amount of investigations have assumed the direction of the load applied to the implant to be in three directions, horizontal, vertical and oblique. Canay *et al.* (1996) and Tada *et al.* (2003) both assumed a horizontal load of 50 N and a vertical load of 100 N with no oblique forces. Whereas Meijer *et al.* (1993) assumed a horizontal force of 10 N, vertical force of 35 N and an oblique force

Table 7. Typical load level (Aparicio and Orozco 1998).

Direction	Specification	Load level (Mean $\pm$ SD) (N)
Vertical	Intrusion force	70 $\pm$ 15
Transverse	Buccal moment	170 $\pm$ 50
	Oral moment	90 $\pm$ 12
	Buccal transverse force at crestal bone margin	21 $\pm$ 6
	Oral transverse force at crestal bone margin	11 $\pm$ 1.5
Mesiodistal	Mesial moment	<52 $\pm$ 26
	Distal moment	35 $\pm$ 17

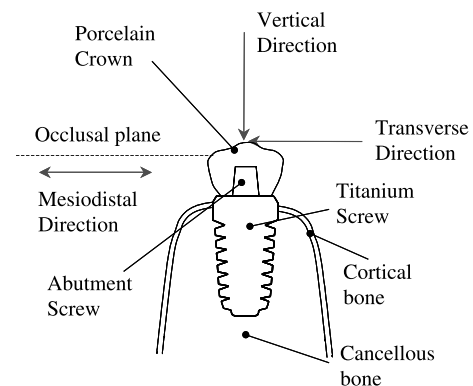


Figure 4. Vertical, transverse and mesiodistal loading directions.



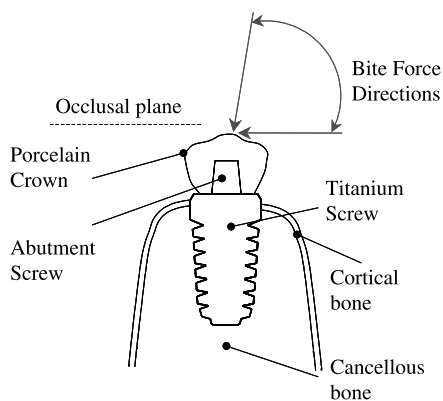


Figure 5. Bite force direction.

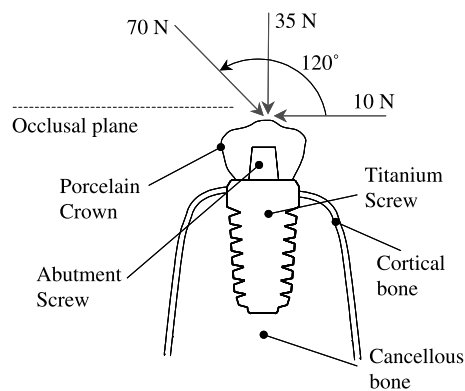


Figure 6. Horizontal, vertical and oblique forces.

of 70 N at an angle of 120° from the horizontal, as shown in figure 6.

Barbier *et al.* (1998) found that the horizontal loading of a freestanding implant evoked the largest maximum stresses, although the magnitude of the applied horizontal load was five times smaller than that of the vertical load. Non-axial loading has often been related to marginal bone loss, failure of osseointegration, failure of the implant and/or the prosthetic components and if connected to natural teeth, failure of the cement seal on the natural tooth (Sullivan 1986, Brunski 1988, Kirsch and Ackermann 1989).

#### 4.5 Implant surface structure

The surface quality of an implant is a combination of physical, chemical, mechanical and morphologic factors. Wennerberg *et al.* (1994) analyzed four groups of implant surface characteristics, these included group A which was HA coated used by OSTEONBOND†, IMZ‡, Micro-Vent¶, Bio-Vent¶, Impla-Med§ and Calcitek||. Group B consisted of titanium plasma-sprayed used by IMZ and group C which was titanium alloy implants used in industry by Core-Vent¶, Screw-Vent¶ and 3I# Miniplant. Finally, group D was commercially pure titanium used in industry by 3I, Impla-Med and Nobelpharma\*\*.

Various aspects of surface quality have been studied, some showing the presence of surface contaminants (although the biological implications of these are unclear) and better bone fixation of rough compared to smooth surfaces. Implant surface characteristics have a dramatic effect on the speed and integrity of osseointegration and the maintenance of soft tissue and surrounding jawbone around the implant. Surface modification is initiated by

factors such as stress reduction and increased surface area. Hermann *et al.* (2001) found that the rough/smooth implant surface has a significant effect on marginal bone formation as evaluated by standardized longitudinal radiography. Bone remodelling occurs rapidly during the early healing phase after implant placement for non-submerged implants and after abutment connection for submerged implants. Figure 7 shows the difference between a submerged and non-submerged implant. Non-submerged implants are common at the time of insertion; whereas submerged implants are expected after 3 months of healing.

The FEM has not been employed widely to evaluate the effect that the surface roughness of the implant has on the stress profile produced within the surrounding jawbone. Ronold *et al.* (2003) experimentally analysed the optimum value for titanium implant roughness in bone attachment using a tensile test. The results supported observations from earlier studies that suggested an optimal surface roughness for bone attachment to be in the range between 3.62 and 3.90 µmm. The analysis also indicated that a further attachment depend on mechanical interlocking between bone and implant.

Wennerberg and Albrektsson (2000) found that an implant with increased surface roughness leads to a more rapid bone response, particularly to the intermediate rough surfaces (1–1.5 µm). The implant surface roughness was determined using the root mean square (RMS) roughness “equation (1)”, as shown below:

$$\text{RMS roughness} = \left\{ \frac{1}{n} \sum_{\chi=0}^n [s(\chi) - \overline{s(\chi)}]^2 \right\}^{0.5} \quad (1)$$

†[http://www.wheelsonline.com/ortho/cement\\_mixing\\_technique\\_for\\_tjr](http://www.wheelsonline.com/ortho/cement_mixing_technique_for_tjr) (accessed 2nd January 2005).

‡[http://www.maryland-implants.com/slavkin\\_implants.htm](http://www.maryland-implants.com/slavkin_implants.htm) (accessed 23rd December 2004).

¶[http://niznick.net/newsletter\\_8.htm](http://niznick.net/newsletter_8.htm) (accessed 12th December 2004).

§<http://www.partisanmgmt.com/implamed/implamed.htm> (accessed 15th January 2005).

||<http://www.calcitek.com/zimmerDental.asp> (accessed 25th February 2005).

#<http://www.3i.com> (accessed 20th November 2004).

\*\*[http://niznick.com/company/1989/cl\\_announcement\\_disclaimer\\_02.htm](http://niznick.com/company/1989/cl_announcement_disclaimer_02.htm) (accessed 4th March 2005).

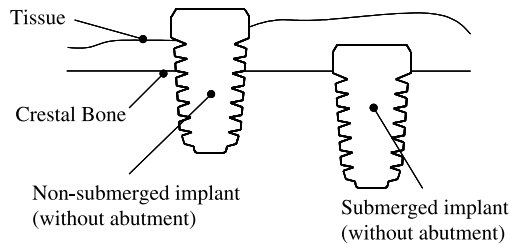


Figure 7. Submerged and non-submerged implants.

where,  $s(x)$  is the surface height at point  $x$  in the surface profile,  $\overline{s(x)}$  is the average height of the surface profile and  $n$  is number of samples collected<sup>†</sup>. These surfaces have in general been compared to the so-called machined or turned surfaces, with little emphasis being paid to the range of surface textures that might exist within a machined screw implant. For crestal bone, Hansson (1999) compared implants with smooth necks to those with screw thread along the entire implant to bone contact surface. It was found that a major decrease in peak interfacial shear stresses occurred if the entire implant to bone contact had a screw thread. It was concluded that the implants with screw thread along the entire surface would counteract marginal bone resorption in accordance with Wolff's law (Wolff 1986). It has been found by various authors (Misch *et al.* 2004, Pelsoczi *et al.* 2004) that the osseointegration of the implant to the surrounding jawbone can be enhanced by simply increasing the surface area of the implant body. Contradictory to Pelsoczi *et al.* (2004) and Misch *et al.* (2004), Rokni *et al.* (2005) found that the implant surface area had insignificant affects on the steady-state crestal bone levels in partially edentulous patients treated with sintered porous-surfaced dental implants.

Threaded implants have an advantage when compared to un-threaded but porous surfaced implants, such as the Endopore<sup>‡</sup> design. With threaded implants, a primary stability is achieved through the thread design however the fixation which occurs at the bone–implant interface, can allow for rotational movement. Whereas a porous design greatly increases the surface area at the interface between bone and implant and hence shorter implants can be used. This, in turn, means that Endopore implants can be used in applications for which much longer, threaded implants would be unsuitable. Clinical, radiographic and histological examinations of the Endopore design, have demonstrated that shorter implants, shorter initial healing periods and simpler surgical techniques can be obtained (Maniatopoulos *et al.* 1986). These results have been demonstrated to be consistent in a clinical setting, where success rates exceed 96%.

## 5. Modelling of jawbone

The primary difficulty in simulating the mechanical behaviour of dental implants is the modelling of human bone tissue and its response to applied mechanical forces. The substantial complexity of the mechanical characteristics of bone and its interaction with implant systems has led to major and often incorrect simplifications made in previous analysis.

Being a self-adaptive material, jawbone needs to be stimulated in order to be retained. A sufficiently low stress would lead to bone loss whereas large stress can cause implant–jawbone interface failure. The ultimate aim during implantation is to achieve the optimum stress profile within the surrounding jawbone. Rieger *et al.* (1990) and O'Mahony *et al.* (2000) both assumed that the target stress level for maximum bone growth occurs at 250 psi (1.72 MPa) levelling off to a control level at 400 psi (2.76 MPa). The stresses depend on both the size and shape of the implant and hence efforts were made to optimise implant geometry to maintain the beneficial stress level in a variety of loading scenarios. Hassler *et al.* (1980) showed that the target compressive stress level for maximum bone growth occurs at 1.8 MPa, levelling off to a control level at 2.8 MPa.

The jawbone experiences stress shielding which is an unpredictable biomechanical response to an applied load. Some theoretical work by Cowin and Hegedus (1976) has dealt with the problem of formulating Wolff's law in a quantitative fashion. In this theory, constitutive equations are developed which predict the remodelling response to a given stress. Stability considerations are invoked to obtain some constraints on the parameters in the constitutive equation.

### 5.1 Material properties of jawbone

The long-term clinical performance of a dental implant is dependant on the preservation of good quality bone surrounding the implant and a well-maintained interface between the biomaterial of the implant and the bone. Clinically, jawbone material properties are categorised into four types. Figure 8 shows that types 1 and 2 offer high Young's modulus, while types 3 and 4 have a thin cortex and low Young's modulus with low trabecular density. Jaffin and Berman (1991) stated that 90% of 1054 implants placed were in types 1–3 bones. Only 3% of these fixtures were lost; of the 10% of the fixtures placed in type 4 bone, 35% failed.

The mandibular is classified into cortical, crestal and trabecular tissue, it is a porous material with a complex microstructure. The quality and quantity of bone surrounding the implant influences the load transfer from implant to the jawbone. With the correct stress level within the jawbone, the correct amount of stress shielding

<sup>†</sup>[www.cce.hw.ac.uk/~gmg/rough\\_surf/surfruf.htm](http://www.cce.hw.ac.uk/~gmg/rough_surf/surfruf.htm) (accessed 29 December 2004).

<sup>‡</sup><http://www.rjsteinerdds.com/html/endopore.htm> (accessed 10th March 2005).

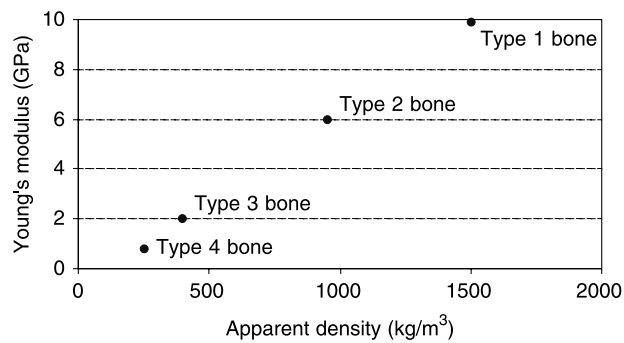


Figure 8. Correlation between Young's modulus and apparent bone densities (Dechow *et al.* 1995).

will occur. Stress shielding is the mechanism that protects the jawbone from stresses that it might encounter by increasing the bone density. Wolff (1986) observed that bone is reshaped in response to the forces acting and this is referred to as Wolff's law. Ideally the selection of the correct implant design, will lead to maximum bone density increase.

Dechow *et al.* (Lotz *et al.* 1991) conducted an investigation into the use of bone plates on the Md in four adult rhesus monkeys. At the beginning and end of the experiments, bone strain was recorded inferior to each bone plate during evoked maximal incisal clenching. After death, gross dimensions and density were measured. An ultrasonic technique was used to measure the material properties, including the elastic and shear moduli. Bone strain inferior to the plates was reduced by 34–53% after attachment of the thick plates. Long-term placement of bone plates and the resulting stress shielding, were found to result in structural changes in the mandibular corpus.

Meijer *et al.* (1993) assumed isotropic, linear elastic and homogenous material properties for bone. But it has been stated by Patra *et al.* (1998) that the material of bone is neither homogeneous nor isotropic and should be modelled as being a porous material with a complex microstructure. It is recognized that cortical bone has

better load bearing capabilities than trabecular bone. Lotz *et al.* (Geng *et al.* 2004) have shown that cortical bone behaves anisotropically, having both transverse and longitudinal module. At the microscopic level, bone is seen to be a composite. The collagen fibres form different arrangements depending on the type of bone. In compact bone, the collagen fibres are very directional. This leads to an orthotropic type material behaviour, such as a classic fibre/matrix composite material. Table 8 shows the assumed material properties for different types of bone used in various studies.

Zarone *et al.* (2003) investigated the biomechanical effect of mandibular functional flexure on stress build-up in implant-supported fixed restorations. Bony tissue was simulated by considering spongy (cancellous) bone to be replaced by an isotropic elastic material, while cortical bone was assumed to be isotropic, with the Young's modulus higher than that of the spongy bone (the physical properties of the materials are reported in table 8). The ratio between spongy and cortical bone thicknesses was approximately 10, as estimated from the autoptic section of the Md; spongy/cortical bone thicknesses were considered constant throughout the mandibular body. It was concluded, during Zarone's study, that a significant amount of stress in the more distal implants and the superstructure at the symphysis arises as a consequence of Md functional flexure. The analysis of the stress distributions generated by the different restorative patterns suggests that a division of the superstructure at the level of the symphysis significantly restores the natural functional flexure of the Md.

## 5.2 Boundary conditions

Most FEA have only considered a small part of the jawbone surrounding the implant (Hobkirk and Schwab 1991, Canay *et al.* 1996, Papavasiliou *et al.* 1997, Patra *et al.* 1998, Pierrisnard *et al.* 2003, Tada *et al.* 2003). It has been found common during the construction of the finite element model, of the jawbone, to apply fixed constraints

Table 8. Assumed material properties for different types of bone.

Source		Papavasiliou <i>et al.</i> (1997)	Pierrisnard <i>et al.</i> (2002)	Patra <i>et al.</i> (1998)	Canay <i>et al.</i> (1996)	Meijer <i>et al.</i> (1993)	Borchers and Reichart (1983)	Zarone <i>et al.</i> (2003)	Lewinstein <i>et al.</i> (1995)	Menicucci <i>et al.</i> (2002)	Farah <i>et al.</i> (1989)	MacGregor <i>et al.</i> (1980)
Cortical bone	Young modulus, $E$ (GPa)	13.7	14	7.5	19.73	13.7	13.7	15	20	13.7	10	*N/A
	Poissons ratio, $\nu$	0.3	0.35	2.7	0.3	0.3	0.3	0.25	0.3	0.3	0.3	*N/A
Cancellous bone	Young modulus, $E$ (GPa)	7.93	2.5	*N/A	*N/A	1.37	*N/A	1.5	*N/A	1.37	*N/A	*N/A
	Poissons ratio, $\nu$	0.3	0.3	*N/A	*N/A	0.3	*N/A	0.29	*N/A	0.3	*N/A	*N/A
Trabecular bone	Young modulus, $E$ (GPa)	*N/A	*N/A	0.5	*N/A	*N/A	*N/A	*N/A	*N/A	*N/A	*N/A	0.25
	Poissons ratio, $\nu$	*N/A	*N/A	0.3	*N/A	*N/A	*N/A	*N/A	*N/A	*N/A	*N/A	0.3

\*N/A, information not available.

to the lower region of the jawbone (Hobkirk and Schwab 1991, Patra *et al.* 1998).

Patra *et al.* (1998) modelled bone support using muscle attachment with spring constraints as opposed to fixed constraints used in previous studies. Ishigaki *et al.* (2003) determined the directions of displacement constraints which were applied to the jawbone, according to the angles of the closing pathways of chopping type (C-type) and grinding type (G-type) chewing patterns. Directions of displacement constraints at the bottom of the Md simulated the calculated directions of the closing pathways for C-type and G-type chewing patterns. The magnitude of the chewing force was selected as 200 N. The sum of the nodal force on the mandibular occlusal surface was 200 N after displacement constraints were applied to the model. The models were restrained at the base of the maxillary first molar to avoid sliding of the entire model.

Ideally, the entire jawbone structure should be evaluated for its contribution to the force exerted unto the dental implant. Hobkirk and Schwab (Lindquist *et al.* 1996) analytically examined the mechanical effect of the contraction of the lateral pterygoid muscles (depresses and protracts Md to open mouth) during opening and protrusive movements of the Md. According to previous studies, the lateral pterygoid muscles pulling medially on the condyles mainly cause flexure of the Md. It was assumed that the effect of these muscles is equivalent to two concentrated loads applied on the condyles along the horizontal rotational condylar axis. An arbitrary load of 10 N was applied in the *x*-direction on the mandibular model in the presence of different typologies of fixed prosthetic restorations.

## 6. Modelling of interface

The periodontal ligament (PDL) in natural teeth provides a spring-like reaction. Creating an implant model with a PDL positioned between the implant and surrounding jawbone, has not been accomplished. Previous research has considered replicating the mechanical effect of the PDL by designing an abutment, which provides a cushioning effect similar to that of the periodontal ligament. This abutment will compress by 0.5 mm when 78.5 N of axial load is applied to the implant (Kaewsuriyathumrong and Soma 1993). In this way, it provides a “buffering” action during the masticatory process to reduce the loading responsible for premature failure of implants. But for the required strength of the abutment screw, it was found that the abutment would become loose after it has been loaded several times.

Kaewsuriyathumrong and Soma (Skalak and Zhao 2000) investigated the stress created by the bite force and which is distributed along the tooth towards the PDL structure. 3D FEA was performed to examine the role of the tooth and PDL structures in the stress distribution. A mandibular first molar was constructed in the finite element model. The bite forces were measured by pressensor and these bite force

values were programmed to load down upon the occlusal surface of the model. When comparing the stress values of sampling points positioned between the root surfaces and the periphery of the PDL (the alveolar wall), all principal stresses for those of the PDL were less than those of the root surfaces. These findings revealed that the PDL, the dentine, and the pulp functioned in cooperation in stress reduction; and the sequences of enamel, dentine and pulp influenced the pattern of stress distribution. The different material properties of the tooth structure in sequence were considered a very important factor for stress reduction and for the pattern of stress distribution, especially in the root. Skalak and Zhao (Stanford and Schneider 2004) stated that close apposition of bone to the titanium implant surface means that under loading, the interface moves as a unit without any relative motion; this is essential for the transmission of stress from the implant to the bone at all parts of the interface.

Stanford and Schneider (Van Rietbergen *et al.* 1993) considered how the material properties of bone on implant surfaces influence biological responses to loading. It was found that the osseous interface of a dental implant responds with a visco-elastic reaction. The titanium oxide layer, on the implant, provides a macroscopic and microscopic architecture which controls the shear strains at the interface.

## 7. Recommendations for future research directions

The various aspects associated with the finite element modelling and analysis of a dental implant and surrounding jawbone have been discussed in some detail in the preceding sections. Extensive research has been completed in the past that has led to significant findings relating to the biomechanical behaviour of the implant, jawbone and their interaction. An in-depth understanding of implant related problems, the development of an optimum implant design and the insertion technique still merit further and advanced investigation.

An initial recommendation involves the construction of a realistic jawbone model. For future research, it might be recommended to identify a wider range of jawbone material properties according to the age and health condition of the patients. A possible method by which today's technology can be employed to perform such a task might include the use of CT image production and computer image processing technique. This technique has been used in hip joint research, (Liu and Ma 1999) however, its application in dentistry is limited. There is a great potential in using CT-based image processing technique in dental implant research, as it is easy to distinguish between the cortical and trabecular bones from the CT images. Once a set of digital CT images of the jawbone are obtained, a precise 3D geometric model of jawbone and surrounding tissue can be readily established using image processing and reverse engineering



approaches (Marco *et al.* 1998, George and Rasha 2001, Wei and Pallavi 2002).

The biomechanical response of the jawbone to a foreign object has not been modelled previously using numerical methods. The jawbone experiences stress shielding which is an unpredictable biomechanical response to an applied load. Ideally, a computer programme/software can be developed which can predict the degree of stress shielding that the jawbone surrounding the implant experiences. The stress shielding produced by numerous jawbone types might also be evaluated through the use of CT-scanned images at set intervals during the healing stage. This will lead to an improved knowledge, for both the implant designer and implant surgeon, of how the jawbone will remodel in different implant placement insertion techniques, designs and loading conditions. An alternative method by which the stress shielding within the jawbone can be evaluated experimentally might be through the use of a photoelastic stress analysis technique (Waskewicz *et al.* 1994).

Numerous investigations have been aimed at determining the optimum geometry of an implant body. Some researchers indicated that the implant geometry had little or no effect on the success of the implant (Ivanoff *et al.* 1999, Pierrisnard *et al.* 2003). While other researchers concluded that an implant diameter of 3.6–4.2 mm provides the highest success rate (Himmlova *et al.* 2004). A new methodology might be employed to determine the optimum combination of length, diameter, taperage and implant cutting flute dimensions for each bone type. It is proposed that such a methodology might involve the use of the Application Programming Interface (API) function of commercial software. This software redevelopment work will lead to a 3D multi-parameter optimum design of the implant based on CAD parameterisation design technique (Vena *et al.* 2000).

When an implant is surgically placed into the jawbone, the implant is mechanically screwed into a drilled hole of a smaller diameter. Large stresses will occur due to the torque applied in the process and the fact that the implant is cutting into the jawbone. As such, the stress condition in the jawbone will change accordingly. This phenomenon has not been researched adequately and the long-term effects of such stresses are still unclear and should therefore be investigated and the solution be sought so that undesirable stresses can be minimised.

The recommended future research areas, discussed above, will aid dentistry by providing an improved understanding of the effect that the implant geometry and orientation has on the stress distribution within the jawbone, during and after the implant is surgically placed into the jawbone. A reduction in the discomfort experienced by the patient during and after implant treatment might occur due to an improved method of surgically placing the implant into the bone.

## 8. Conclusion

FEA has been used extensively to predict the biomechanical performance of various dental implant designs as well as the effect of clinical factors on the success of implantation. The principal difficulty in simulating the mechanical behaviour of dental implants is the modelling of the living human bone tissue and its response to applied mechanical forces. Research has been conducted on the design philosophy, length, diameter and shape of implants as well as the biomechanical bond formed between the implant and the jawbone. Such work has been mainly directed towards finding the most biocompatible materials with which the dental implants are constructed, in an attempt to reduce the potential risks of clinical failure. Despite the numerous research efforts in the field, no ideal design philosophy is available and the biological interaction at the interface between the implant material and the living tissue is not yet fully understood.

An in-depth understanding of stress profiles encountered by the implant and more importantly in the surrounding jawbone can be gained through the use of FEM. This increase in knowledge of stress distributions and magnitudes within the implant and surrounding jawbone will aid the optimisation of the implant design and insertion technique. It is of great importance that the clinician gains an understanding of the methodology, applications and limitations of FEA in implant dentistry and become more confident to interpret results of FEA studies and interpret these results 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 goes to Mr John Mackay and Ms Jane Evans for their endless contribution.

## References

- C. Aparicio and P. Orozco, "Use of 5-mm-diameter implants: periosteal values related to a clinical and radiographic evaluation", *Clin. Oral Implants Res.*, 9(6), pp. 398–406, 1998.
- K. Akca and H. Iplikcioglu, "Finite element stress analysis of the influence of staggered versus straight placement of dental implants", *Int. J. Oral Maxillofac. Implants*, 16(5), pp. 722–730, 2001.
- K. Akca, M.C. Cehreli and H.A. Iplikcioglu, "Comparison of three-dimensional finite element stress analysis with *in vitro* strain gauge measurements on dental implants", *Int. J. Prosthodont.*, 15(2), pp. 115–121, 2002.
- I. Akpinar, N. Anil and L. Parnas, "A natural tooth's stress distribution in occlusion with a dental implant", *J. Oral Rehabil.*, 27(6), pp. 538–545, 2000.
- P. Astrand, C. Billstrom, H. Feldmann, K. Fischer, V. Henriksson, B. Johansson, E. Nystrom and B. Sunzel, "Tapered implants in jaws with soft bone quality: a clinical and radiographic 1-year study of the Branemark System Mark IV fixture", *Clin. Implant Dent. Relat. Res.*, 5(4), pp. 213–218, 2003.
- L. Barbier, J. Vander Sloten, G. Krzesinski, E. Schepers and G. Van der Perre, "Finite element analysis of non-axial versus axial loading of oral implants in the mandible of the dog", *J. Oral Rehabil.*, 25(11), pp. 847–858, 1998.



- L. Borchers and P. Reichart, "Three-dimensional stress distribution around a dental implant at different stages of interface development", *J. Dent. Res.*, 62(2), pp. 155–159, 1983.
- J.B. Brunski, "Biomechanics of oral implants: future research directions", *J. Dent. Educ.*, 52(12), pp. 775–787, 1988.
- S. Canay, N. Hersek, I. Akpinar and Z. Asik, "Comparison of stress distribution around vertical and angled implants with finite-element analysis", *Quintessence Int.*, 27(9), pp. 591–598, 1996.
- S.C. Cowin and D.H. Hegedus, "Bone remodeling I: theory of adaptive elasticity", *J. Elasticity*, 6, pp. 313–326, 1976.
- D.T. Davy, G.L. Dilley and R.F. Krejci, "Determination of stress patterns in root-filled teeth incorporating various dowel designs", *J. Dent. Res.*, 60(7), pp. 1301–1310, 1981.
- P.C. Dechow, E. Ellis and G.S. Throckmorton, "Structural properties of mandibular bone following application of a bone plate", *J. Oral Maxillofac. Surg.*, 53(9), pp. 1044–1051, 1995.
- D.H. DeTolla, S. Andreana, A. Patra, R. Buhite and B. Comella, "Role of the finite element model in dental implants", *J. Oral Implantol.*, 26(2), pp. 77–81, 2000.
- I. Ergatoudis, B.M. Irons and O.C. Zienkiewicz, "Curved, isoparametric, quadrilateral elements for finite element analysis", *Int. J. Solids Struct.*, 4(1), pp. 31–42, 1968.
- G. Eskitascioglu, A. Usumez, M. Sevimay, E. Soykan and E. Unsal, "The influence of occlusal loading location on stresses transferred to implant-supported prostheses and supporting bone: a three-dimensional finite element study", *J. Prosthet. Dent.*, 91(2), pp. 144–150, 2004.
- J.W. Farah, J.A. Hood and R.G. Craig, "Stresses and deflections in the floor of model cavity preparations", *J. Oral Rehabil.*, 1(2), pp. 207–215, 1974.
- J.W. Farah, R.G. Craig and K.A. Meroueh, "Finite element analysis of three- and four-unit bridges", *J. Oral Rehabil.*, 16(6), pp. 603–611, 1989.
- B. Friberg, S. Jisander, G. Widmark, A. Lundgren, C.J. Ivanoff, L. Sennerby and C. Thoren, "One-year prospective three-center study comparing the outcome of a "soft bone implant" (prototype Mk IV) and the standard Branemark implant", *Clin. Implant Dent. Relat. Res.*, 5(2), pp. 71–77, 2003.
- J.P. Geng, K.B. Tan and G.R. Liu, "Application of finite element analysis in implant dentistry: a review of the literature", *J. Prosthet. Dent.*, 85(6), pp. 585–598, 2001.
- J.P. Geng, Q.S. Ma, W. Xu, K.B.C. Tan and G.R. Liu, "Finite element analysis of four thread-form configurations in a stepped screw implant", *J. Oral Rehabil.*, 31(3), pp. 233–239, 2004.
- K.K. George and A.N. Rasha, "Adaptive reconstruction of bone geometry from serial cross-sections", *Artif. Intell. Eng.*, 15, pp. 227–239, 2001.
- S. Hansson, "The implant neck: smooth or provided with retention elements. A biomechanical approach", *Clin. Oral Implants Res.*, 10(5), pp. 394–405, 1999.
- T. Haraldson, T. Jemt, P.A. Stalblad and U. Lekholm, "Oral function in subjects with overdentures supported by osseointegrated implants", *Scand. J. Dent. Res.*, 96(3), pp. 235–242, 1988.
- C.R. Hassler, E.F. Rybicki, K.D. Cummings and L.C. Clark, "Quantification of bone stresses during remodeling", *J. Biomech.*, 13(2), pp. 185–190, 1980.
- H.S. Hedia and N.A. Mahmoud, "Design optimization of functionally graded dental implant", *Biomed. Mater. Eng.*, 14(2), pp. 133–143, 2004.
- J.S. Hermann, J.D. Schoolfield, P.V. Nummikoski, D. Buser, R.K. Schenk and D.L. Cochran, "Crestal bone changes around titanium implants: a methodologic study comparing linear radiographic with histometric measurements", *Int. J. Oral Maxillofac. Implants*, 16(4), pp. 475–485, 2001.
- L. Himmlova, T. Dostalova, A. Kacovsky and S. Konvickova, "Influence of implant length and diameter on stress distribution: a finite element analysis", *J. Prosthet. Dent.*, 91(1), pp. 20–25, 2004.
- J.A. Hobkirk and J. Schwab, "Mandibular deformation in subjects with osseointegrated implants", *Int. J. Oral Maxillofac. Implants*, 6(3), pp. 319–328, 1991.
- E.P. Holmgren, R.J. Seckinger, L.M. Kilgren and F. Mante, "Evaluating parameters of osseointegrated dental implants using finite element analysis—a two-dimensional comparative study examining the effects of implant diameter, implant shape, and load direction", *J. Oral Implantol.*, 24(2), pp. 80–88, 1998.
- H. Iplikcioglu and K. Akca, "Comparative evaluation of the effect of diameter, length and number of implants supporting three-unit fixed partial prostheses on stress distribution in the bone", *J. Dent.*, 30(1), pp. 41–46, 2002.
- S. Ishigaki, T. Nakano, S. Yamada, T. Nakamura and F. Takashima, "Biomechanical stress in bone surrounding an implant under simulated chewing", *Clin. Oral Implants Res.*, 14(1), pp. 97–102, 2003.
- Y.H. Ismail, L.N. Pahountis and J.F. Fleming, "Comparison of two-dimensional and three-dimensional finite element analysis of a blade implant", *Int. J. Oral Implantol.*, 4, pp. 25–31, 1987.
- C.J. Ivanoff, K. Grondahl, L. Sennerby, C. Bergstrom and U. Lekholm, "Influence of variations in implant diameters: a 3–5-year retrospective clinical report", *Int. J. Oral Maxillofac. Implants*, 14(2), pp. 173–180, 1999.
- R.A. Jaffin and C.L. Berman, "The excessive loss of Branemark fixtures in type IV bone: a 5-year analysis", *J. Periodontol.*, 62(1), pp. 2–4, 1991.
- C. Kaewsuriyathumrong and K. Soma, "Stress of tooth and PDL structure created by bite force", *Bull. Tokyo Med. Dent. Univ.*, 40(4), pp. 217–232, 1993.
- A. Kirsch and K.L. Ackermann, "The IMZ osteointegrated implant system", *Dent. Clin. North Am.*, 33(4), pp. 733–791, 1989.
- R.J. Kohal, G. Papavasiliou, P. Kamposiora, A. Tripodakis and J.R. Strub, "Three-dimensional computerized stress analysis of commercially pure titanium and yttrium-partially stabilized zirconia implants", *Int. J. Prosthodont.*, 15(2), pp. 189–194, 2002.
- J.H. Koolstra, T.M. van Eijden, W.A. Weijs and M. Naeije, "A three-dimensional mathematical model of the human masticatory system predicting maximum possible bite forces", *J. Biomech.*, 21(7), pp. 563–576, 1988.
- I. Lewinstein, L. Banks-Sills and R. Eliasi, "A finite element analysis of a new system (IL) for supporting an implant-retained cantilever prosthesis", *Int. J. Oral Maxillofac. Implants*, 10(3), pp. 355–366, 1995.
- L.W. Lindquist, G.E. Carlsson and T. Jemt, "A prospective 15year followup study of mandibular fixed prostheses supported by osseointegrated implants", *Clin. Oral Implants Res.*, 7(4), p. 329, 1996.
- S. Liu and W. Ma, "Seed-growing segmentation of 3-D surfaces from CT-contour data", *Comput. Aided Des.*, 31, pp. 517–536, 1999.
- J.C. Lotz, T.N. Gerhart and W.C. Hayes, "Mechanical properties of metaphyseal bone in the proximal femur", *J. Biomech.*, 24(5), pp. 317–329, 1991.
- A.R. MacGregor, T.P. Miller and J.W. Farah, "Stress analysis of mandibular partial dentures with bounded and free-end saddles", *J. Dent.*, 8(1), pp. 27–34, 1980.
- G. Mailath, B. Stoiber, G. Watzek and M. Matejka, "Bone resorption at the entry of osseointegrated implants—a biomechanical phenomenon. Finite element study", *J. Oral Implantol.*, 24(2), pp. 80–88, 1989.
- C. Maniopoulos, R.M. Pilliar and D.C. Smith, "Threaded versus porous-surfaced designs for implant stabilization in bone-endodontic implant model", *J. Biomed. Mater. Res.*, 20(9), pp. 1309–1333, 1986.
- V. Marco, Z. Cinzia and P. Luisa, "TRISOLID: an application of reverse engineering methods to the creation of CAD models of bone segments", *Comput. Methods Programs Biomed.*, 56, pp. 211–220, 1998.
- H.J. Meijer, F.J. Starman, W.H. Steen and F. Bosman, "A three-dimensional, finite-element analysis of bone around dental implants in an edentulous human mandible", *Arch. Oral Biol.*, 38(6), pp. 491–496, 1993.
- G. Menicucci, A. Mossolov, M. Mozzati, M. Lorenzetti and G. Preti, "Tooth-implant connection: some biomechanical aspects based on finite element analyses", *Clin. Oral Implants Res.*, 13(3), pp. 334–341, 2002.
- N. Meredith, F. Shagaldi, D. Alleyne, L. Sennerby and P. Cawley, "The application of resonance frequency measurements to study the stability of titanium implants during healing in the rabbit tibia", *Clin. Oral Implants Res.*, 8(3), pp. 234–243, 1997.
- W.M. Mihalko, T.C. May, J.F. Kay and W.R. Krause, "Finite element analysis of interface geometry effects on the crestal bone surrounding a dental implant", *J. Implant Dent.*, 1(3), pp. 212–217, 1992.
- C.E. Misch, "Implant design considerations for the posterior regions of the mouth", *Implant Dent.*, 8(4), pp. 376–386, 1999.
- C.E. Misch, H.L. Wang, C.M. Misch, M. Sharawy, J. Lemons and K.W. Judy, "Rationale for the application of immediate load in implant dentistry: part II", *Implant Dent.*, 13(4), pp. 310–321, 2004.

- H.H. Moroi, K. Okimoto, R. Moroi and Y. Terada, "Numeric approach to the biomechanical analysis of thermal effects in coated implants", *Int. J. Prosthodont.*, 6(6), pp. 564–572, 1993.
- T. Nagasao, M. Kobayashi, Y. Tsuchiya, T. Kaneko and T. Nakajima, "Finite element analysis of the stresses around endosseous implants in various reconstructed mandible models", *J. Craniomaxillofac. Surg.*, 30(3), pp. 170–177, 2002.
- T. Nagasao, M. Kobayashi, Y. Tsuchiya, T. Kaneko and T. Nakajima, "Finite element analysis of the stresses around fixtures in various reconstructed mandibular models—part II (effect of horizontal load)", *J. Craniomaxillofac. Surg.*, 31(3), pp. 168–175, 2003.
- K. Nishihara and S. Nakagiri, "Biomechanical studies on newly tailored artificial dental root", *Biomed. Mater. Eng.*, 4(3), pp. 141–149, 1994.
- Nobel Biocare, Available online at: [www.nobelbiocare.com](http://www.nobelbiocare.com) (accessed 10 December 2004).
- W.J. O'Brien, *Dental Materials: Properties and Selection*, Chicago; London: (ISBN 0867151994) Quintessence Publishing, 1989.
- A. O'Mahony, Q. Bowles, G. Woolsey, S.J. Robinson and P. Spencer, "Stress distribution in the single-unit osseointegrated dental implant: finite element analyses of axial and off-axial loading", *Implant Dent.*, 9(3), pp. 207–218, 2000.
- D. O'Sullivan, L. Sennerby and N. Meredith, "Influence of implant taper on the primary and secondary stability of osseointegrated titanium implants", *Clin. Oral Implants Res.*, 15(4), pp. 474–480, 2004.
- Osstel, Available online at: [www.integrationdiagnostics.com/pro/pro\\_range.html](http://www.integrationdiagnostics.com/pro/pro_range.html) (accessed 29 December 2004)
- G. Papavasiliou, P. Kamposiora, S.C. Bayne and D.A. Felton, "3D-FEA of osseointegration percentages and patterns on implant-bone interfacial stresses", *J. Dent.*, 25(6), pp. 485–491, 1997.
- A.K. Patra, J.M. DePaolo, K.S. D'Souza, D. DeTolla and M.A. Meenaghan, "Guidelines for analysis and redesign of dental implants", *Implant Dent.*, 7(4), pp. 355–368, 1998.
- K.I. Pelsoczi, M. Berezna, Z. Toth, K. Turzo, M. Radnai, Z. Bor and A. Fazekas, "Surface modifications of titanium implant material with excimer laser for more effective osseointegration", *Fogorv. Sz.*, 97(6), pp. 231–237, 2004.
- L. Pierrisnard, G. Hure, M. Barquins and D. Chappard, "Two dental implants designed for immediate loading: a finite element analysis", *Int. J. Oral Maxillofac. Implants*, 17(3), pp. 353–362, 2002.
- L. Pierrisnard, F. Renouard, P. Renault and M. Barquins, "Influence of implant length and bicortical anchorage on implant stress distribution", *Clin. Implant Dent. Relat. Res.*, 5(4), pp. 254–262, 2003.
- H. Plikciolu and K. Akca, "Comparative evaluation of the effect of diameter, length and number of implants supporting three-unit fixed partial prostheses on stress distribution in the bone", *J. Dent.*, 30(1), pp. 41–46, 2002.
- R.A. Reinhardt, Y.C. Pao and R.F. Krejci, "Periodontal ligament stresses in the initiation of occlusal traumatism", *J. Periodontal Res.*, 19(3), pp. 238–246, 1984.
- J.Y. Rho, R.B. Ashman and C.H. Turner, "Young's modulus of trabecular and cortical bone material: ultrasonic and microtensile measurements", *J. Biomech.*, 26(2), pp. 111–119, 1993.
- E.J. Richter, "In vivo horizontal bending moments on implants", *Int. J. Oral Maxillofac. Implants*, 13(2), pp. 232–244, 1998.
- R. Rieger, M. Mayberry and M.O. Brose, "Finite element analysis of six endosseous implants", *J. Prosthet. Dent.*, 63(6), pp. 671–676, 1990.
- S. Rokni, R. Todescan, P. Watson, M. Pharoah, A.O. Adegbenbo and D. Deporter, "An assessment of crown-to-root ratios with short sintered porous-surfaced implants supporting prostheses in partially edentulous patients", *Int. J. Oral Maxillofac. Implants*, 20(1), pp. 69–76, 2005.
- H.J. Ronold, S.P. Lyngstadaas and J.E. Ellingsen, "Analysing the optimal value for titanium implant roughness in bone attachment using a tensile test", *Biomaterials*, 24(25), pp. 4559–4564, 2003.
- J.S. Rzemieniecki, "Theory of matrix structural analysis", *J. Sound Vibration*, 10(2), pp. 358–359, 1969.
- S.W. Shin, S.R. Bryant and G.A. Zarb, "A retrospective study on the treatment outcome of wide-bodied implants", *Int. J. Prosthodont.*, 17(1), pp. 52–58, 2004.
- D. Siegele and U. Soltesz, "Numerical investigations of the influence of implant shape on stress distribution in the jaw bone", *Int. J. Oral Maxillofac. Implants*, 4(4), pp. 333–340, 1989.
- R. Skalak and Y. Zhao, "Similarity of stress distribution in bone for various implant surface roughness heights of similar form", *Clin. Implant Dent. Relat. Res.*, 2(4), pp. 225–230, 2000.
- C.M. Stanford and G.B. Schneider, "Functional behaviour of bone around dental implants", *Gerodontology*, 21(2), pp. 71–77, 2004.
- R. Stegaroiu, H. Kusakari, S. Nishiyama and O. Miyakawa, "Influence of prosthesis material on stress distribution in bone and implant: a 3-dimensional finite element analysis", *Int. J. Oral Maxillofac. Implants*, 13(6), pp. 781–790, 1998.
- D.Y. Sullivan, "Prosthetic considerations for the utilization of osseointegrated fixtures in the partially edentulous arch", *Int. J. Oral Maxillofac. Implants*, 1(1), pp. 39–45, 1986.
- S. Tada, R. Stegaroiu, E. Kitamura, O. Miyakawa and H. Kusakari, "Influence of implant design and bone quality on stress/strain distribution in bone around implants: a 3-dimensional finite element analysis", *Int. J. Oral Maxillofac. Implants*, 18(3), pp. 357–368, 2003.
- G. Tawil, M. Mawla and J. Gottlow, "Clinical and radiographic evaluation of the 5-mm diameter regular-platform Branemark fixture: 2–5-year follow-up", *Clin. Implant Dent. Relat. Res.*, 4(1), pp. 16–26, 2002.
- B. van Rietbergen, R. Huiskes, H. Weinans, D.R. Sumner, T.M. Turner and J.O. Galante, "The mechanism of bone remodeling and resorption around press-fitted THA stems", *J. Biomech.*, 26, pp. 369–382, 1993.
- P. Vena, N. Verdonchot, R. Contro and R. Huiskes, "Sensitivity analysis and optimal shape design for bone-prosthesis interfaces in a femoral head surface replacement", *Comput. Methods Biomech. Biomed. Eng.*, 3(3), pp. 245–256, 2000.
- V.J. Waskewicz, G.A. Ostrowski and J.S. Parks, "Photoelastic analysis of stress distribution transmitted from a fixed prosthesis attached to osseointegrated implants", *J. Oral Maxillofac. Implants*, 9(4), pp. 405–411, 1994.
- F. Watanabe, Y. Hata, S. Komatsu, T.C. Ramos and H. Fukuda, "Finite element analysis of the influence of implant inclination, loading position, and load direction on stress distribution", *Odontology*, 9(1), pp. 31–36, 2003.
- S. Wei and L. Pallavi, "Recent development on computer aided tissue engineering—a review", *Comput. Methods Programs Biomed.*, 67, pp. 85–103, 2002.
- A. Wennerberg and T. Albrektsson, "Suggested guidelines for the topographic evaluation of implant surfaces", *Int. J. Oral Maxillofac. Implants*, 15(3), pp. 331–344, 2000.
- B. Wennerberg, A. Albrektsson and T. Andersson, "Design and surface characteristics of 13 commercially available oral implant systems", *Int. J. Oral Maxillofac. Implants*, 8(6), pp. 622–633, 1994.
- J. Wolff, *The Law of Bone Remodelling*, Berlin: Springer, 1986, p. 1.
- K.W. Wright and A.L. Yettram, "Reactive force distributions for teeth when loaded singly and when used as fixed partial denture abutments", *J. Prosthet. Dent.*, 42(4), pp. 411–416, 1979.
- S. Yokoyama, N. Wakabayashi, M. Shiota and T. Ohyama, "The influence of implant location and length on stress distribution for three-unit implant-supported posterior cantilever fixed partial dentures", *J. Prosthet. Dent.*, 91(3), pp. 234–240, 2004.
- F. Zarone, A. Apicella, L. Nicolais, R. Aversa and R. Sorrentino, "Mandibular flexure and stress build-up in mandibular full-arch fixed prostheses supported by osseointegrated implants", *Clin. Oral Implants Res.*, 14(1), pp. 103–114, 2003.
- L. Zhang, Y. Zhou and W. Meng, "A three-dimensional finite element analysis of the correlation between lengths and diameters of the implants of fixed bridges with proper stress distribution", *Hua Xi Kou Qiang Yi Xue Za Zhi*, 18(4), pp. 229–231, 2000.