John B. Brunski

Biomechanical aspects of the optimal number of implants to carry a cross-arch full restoration

John B. Brunski, PhD
Senior Research Engineer, Division of Plastic and Reconstructive Surgery, Department of Surgery, School of Medicine, Stanford University, Stanford, CA, USA

Correspondence to:
John B. Brunski, PhD
Senior Research Engineer, Division of Plastic and Reconstructive Surgery, Department of Surgery, School of Medicine, PSRL, 257 Campus Drive, Stanford University, Stanford, CA 94305
USA
Mob: 518.229.6963
Email: brunsj6@stanford.edu

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A proper definition of the 'optimal' number of implants to support a full arch prosthesis should go beyond solely a listing of the number of implants used in a treatment plan; it should be based upon a biomechanical analysis that takes into account several factors: the locations of the implants in the jaw; the quality and quantity of bone into which they are placed; the loads (forces and moments) that develop on the implants; the magnitudes of stress and strain that develop in the interfacial bone as well as in the implants and prosthesis; and the relationship of the stresses and strains to limits for the materials involved. Overall, determining an 'optimal' number of implants to use in a patient is a biomechanical design problem.
This paper discusses some of the approaches that are already available to aid biomechanically focused clinical treatment planning. A number of examples are presented to illustrate how relatively simple biomechanical analyses – e.g. the Skalak model – as well as more complex analyses (e.g. finite element modelling) can be used to assess the pros and cons of various arrangements of implants to support full-arch prostheses. Some of the examples considered include the use of 4 rather than 6 implants to span the same arc-length in a jaw, and the pros and cons of using tilted implants as in the 'all-on-4' approach. In evaluating the accuracy of the various biomechanical analyses, it is clear that our current prediction methods are not always perfectly accurate in vivo, although they can provide a reasonably approximate analysis of a treatment plan in many situations. In the current era of cone beam computerised tomography (CT) scans of patients in the dental office, there is significant promise for finite element analyses (FEA) based on anatomically-accurate input data. However, at the same time it has to be recognised that effective use of FEA software requires a reasonable engineering background, especially insofar as interpretations of the clinical significance of stresses and strains in bone and prosthetic materials.

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■ Introduction

This article presents basic biomechanical analyses to guide the optimal use of oral implants in full-arch prosthetic restorations. However, at the outset, the adjective 'optimal' requires some explanation. Definitions of 'optimal' include the following:

- "Most favourable or desirable" (Anonymous, 2009)[1].
- "In mathematics, an optimal solution is one that is determined to be the best solution from all feasible solutions. In business, it is a solution that best fits a situation by employing organizational resources in the most effective and efficient manner" (Anonymous, 2014)[2].

- "[Optimal] describes a solution to a problem which minimizes some cost function" (Howe, 2010)[3].

But what are the criteria for determining what is 'favourable' or 'best'? In searching for an 'optimal' solution to a problem – such as the problem of selecting how many implants are 'best' in treating a full-arch reconstruction of an edentulous jaw – it is important to have some criteria for defining 'optimality'.

In this regard, the last definition above is helpful because it explains that an 'optimal' solution is one that "…minimises some cost function". For instance, 'cost function' can be used in the context of economics: a 'cost function' is an equation (function) whose value depends on several variables ('inputs'), each of which have 'prices' or 'costs'; ultimately this cost function explains how the cost to produce a certain 'output' depends on the prices of the 'inputs'[4]. At least in this economics example, it would be considered 'optimal' to minimise (as opposed to maximise) the 'cost function', since typically when producing goods in a business, it is desirable to reduce the costs of production.

In analogy with the above idea, one way to apply a 'cost function' in patient treatment would be to recognise that a non-economic 'cost' to a patient includes pain and discomfort, limited function, and time of disablement. And while choosing the 'optimal' number of implants to treat an edentulous jaw is not (solely) an economics problem, nevertheless it is instructive to consider how a 'cost function' might be developed to help guide optimal treatment planning with implants. For example, first it would be possible to define a 'risk function' where this function would depend on key 'inputs' (variables) in the problem, including: the number of implants; location of implants in the arch; shape/size/biomaterial of the implants; quality of bone at the implant sites; expected masticatory loading; prosthesis design; loading paradigm (i.e. immediate vs. delayed loading); patient pain and discomfort; patient inconvenience, etc. Second, one could then develop an 'optimal' solution by minimising the defined risk function. This could be done by assigning a 'risk value' to each variable or 'input' in the 'risk function' (analogous to assigning a 'price' to each input in the cost function noted earlier), and then minimising the total risk function with respect to the inputs. Alternatively, it would be possible to take a different approach and develop a 'probability of success function' (PSF), which would also depend on the same variables noted earlier in the risk function, except that here with the PSF, one would seek to maximise this PSF.

In any case, whether dealing with a 'risk function' (or its inverse, a 'probability of success function') defining an optimal therapy with implants involves many inputs (variables, factors) that influence the outcome. Therefore, with intraoral implants, any attempt to define 'optimality' only in terms of the number of implants is incomplete and risks missing the main point – which is that optimality of the treatment depends on more than just the number of implants. While certainly the number of implants is a key factor, so are the length and diameter of the implant(s), how and where the implants are placed in the bone, what the bone properties are, what the prosthesis is made of, how the prosthesis is designed and loaded, whether one is planning for immediate loading or delayed loading, and many non-biomechanical factors such as patient discomfort and related issues, etc.

So in this context, this article answers three main questions that can help define optimality in a biomechanical sense:

1. How does one predict the forces and moments on implants supporting a cross-arch prosthesis *in vivo*?
2. How do certain variables influence the forces and moments on implants, namely, variables including: number of implants; location of implants; length and diameter of implants; length of a cantilever; 'upright' vs. 'tilted' implants; stiffness of the implant in the bone; type of prosthesis, etc.
3. How accurate are our existing methods for predicting the loadings on implants *in vivo*?

This paper will not delve into the clinical evidence about how many implants can or should be used to support full-arch reconstructions; such clinical information is covered in other papers in this issue of the journal. Instead, this article summarises the biomechanical background that can be used to quantitatively evaluate the pros and cons of various ways that clinicians may place implants in full-arch treatments. For more in-depth biomechanical background, it may be useful for readers to consult previous articles related to this topic[5,6].

Biomechanical approach to treatment planning

Before considering detailed calculations and numerical analyses about numbers of implants etc., it is important to consider the over-arching design perspective surrounding treatment planning with oral implants. When designing any load-bearing structure, a primary goal is to design against mechanical failure, in all of the ways in which mechanical failure might manifest itself. Depending on the nature of the structure being considered and how it will be loaded, mechanical failure is possible by a number of mechanisms, such as single-cycle overload, fatigue under cyclic loading condition, yielding, etc. In the specific instance of treatment planning with oral implants, a flowchart (Fig 1) helps to illustrate a step-by-step design paradigm by which biomechanical case planning can unfold. It's easy to imagine that the steps in this flowchart could be applicable to many common mechanical design problems, including, for example, deciding how best – from architectural and structural viewpoints – to build a small wooden deck behind a house, or how best to design a large skyscraper. The mention of architectural and structural design is apt because it suggests the importance of defining and adhering to certain well-accepted quantitative 'building codes' to assure a safe and effective construction. Indeed, building codes in the construction industry have – or should have – analogues when it comes to design and construction of any full-arch implant-supported prosthesis to restore a mandible or maxilla.

So, consider Step 1 of the treatment planning algorithm (Fig 1): a clinician starts to consider factors such as the patent's oral health history, bone of the dental arches, what the prosthesis might look like, how many implants might be used, where those implants might be placed to support the prosthesis, what sorts of implants might be used, and what sorts of functional loading are likely in this patient.

Then Step 2 is to analyse the biomechanics in more detail, based on the initial plans conceived in

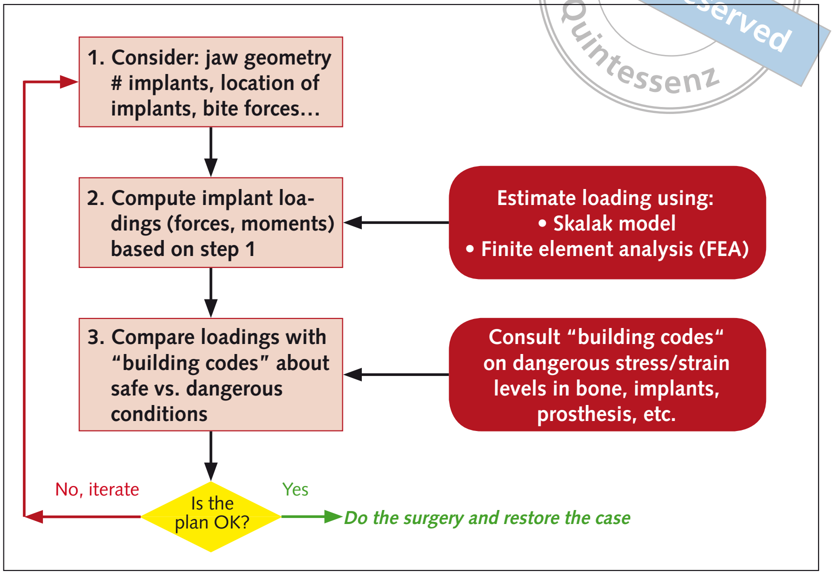


Fig 1 Schematic biomechanical paradigm for treatment planning.

Step 1. For example, if 6 implants are initially contemplated to support a full-arch mandibular denture in a delayed loading scenario, then the clinician would start to estimate what loadings (forces and moments) are anticipated on the implants, and how such loadings would factor into whether the implant performance will be optimal. (This can be done by several methods to be discussed shortly.) Among the numerous factors influencing this analysis is how the 6 implants are situated relative to one another in the jaw, the arc-length over which they are spread, the size/material of the prosthesis, and how the prosthesis will be loaded. For example, in one tentative plan, 6 implants might be equally spaced between the mental foramina in the mandible, whereas in another possible treatment plan, 4 implants might be spaced over that same arc-length. And perhaps in each plan the distal cantilever lengths are, say, 20 mm, and the largest biting forces on the proposed prosthesis occur at those distal locations. A clinician might want to consider several possible plans, but in any event, each plan would be examined further using calculations about the loadings on the implants. Finally, at the end of this Step 2, the main outcome would consist of quantitative results about the anticipated forces and moments on each implant in each of the possible treatment plans that have been considered.

In Step 3, the clinician would take the results from Step 2 and make more detailed analyses about the significance of the loadings on each implant.

For example, for various reasons besides just the biomechanics at this stage – perhaps economic considerations, or issues of bone quantity in certain locations in a patient's maxilla – the clinician might decide to more closely examine a treatment plan involving just 4 implants to avoid the 'cost' of bone augmentation procedures. Then one key analysis that needs to occur involves answering the following question: Suppose the analysis in Step 2 reveals that a 3.75 mm diameter × 10 mm long implant in the plan with 4 implants will experience an axial compressive force of 250 N and a mesiodistal bending moment of 20 N-cm: is this loading going to create improper levels of stress and strain in the bone around this implant? (Actually, this question has to be answered for each implant, since the loading of each implant in a distribution is not going to be the same, as will be clear from some examples to be considered shortly.)

In principle, an answer to this stress-analysis question may seem straightforward. Look up the stress-strain limits for interfacial bone and compare them to the predicted stress-strain levels found in our analysis in Step 3; then, if the predicted stress-strain levels exceed certain limits, reconsider the original plan so as to reduce to proper levels the loadings on implant(s) in question. Unfortunately, at this stage of our understanding of implants and interfacial bone, getting a satisfactory answer to this central stress-analysis question remains problematic. (Indeed, Step 3 is not practiced by clinicians, although as research continues, this step will likely become more practical, if not routine.) The reason for Step 3's difficulty is that the ability to make accurate predictions of the stress-strain levels in bone around oral implants – for instance using finite element (FE) computational models – requires accurate input data that is not always available, e.g. 1) the quantity and spatial location of interfacial bone; 2) the exact mechanical properties of that interfacial bone (e.g. its elastic modulus, stress-strain limits in terms of ultimate, yield, and fatigue strengths); and quantitative rules describing bone's long-term modelling/remodelling response to interfacial stress-strain conditions. While the technical capability of modern commercial finite element (FE) software is outstanding, a relevant programmer's adage still applies: 'GIGO, Garbage In, Garbage Out'. So while progress is being made in these types of interfacial stress analyses – and certainly a rudimentary level of stress analysis is possible – the unfortunate fact is that the oral implant field currently lacks a robust set of 'building codes' for making fully accurate, clinically and biologically reliable assessments of interfacial stresses and strains around oral implants. To make a comparison, if the predictive success of stress analysis were to be ranked on a 1 to 10 scale, with 10 being excellent and 1 being poor, the predictive success of analyses used in designing common engineering structures such as modern buildings and jet engines would be at a 9 or 10, while the validity of analyses used in assessing bone around oral implants would be at a 6 or 7.

After Step 3, the treatment planning process reaches a cautionary decision box (Fig 1) asking, 'Is the plan OK?' If the answer is 'No', the algorithm reverts back to Step 1, for a redesign effort that could involve the use of more implants or different implant locations, or perhaps wider or longer implants, or perhaps a different prosthesis design, etc. On the other hand, if after Step 3 the treatment plan looks 'OK', then the clinician continues with the rest of the planning, with a focus on the remaining steps, e.g. details of the surgery, prosthetics, etc.

Two additional points are useful in the context of Step 3 and the associated stress-strain analysis alluded to in the oval to the right in Fig 1. First, if stresses or strains become too large in a material in a structure, the material will fail, compromising the integrity of the structure. Obviously materials can fail mechanically in a number of ways, such as by yielding, fracture, or fatigue. (A summary of the basics of mechanical failure appears in other references[7].) Second, the stresses and strains in materials in a structure depend on the external loads that act on the structure. So in deciding on how many implants will properly support a prosthesis, the designer must also know as much as possible about the external loadings on the prosthesis, implant, and interfacial bone. As will be seen especially in the example of using 'upright' versus 'tilted' implants to support a prosthesis in an 'all-on-4' system, this issue of loading (as well as stresses and strains in the implants, prosthesis, and interfacial bone) becomes decisive.

Methods to predict loading on oral implants

Methods for analysing the forces and moments on oral implants have been discussed in several previous publications and textbook chapters[5,8,9]. When implants support a prosthesis, each implant must act – in the language of basic mechanics – as a 'fixed connection'. This means that each implant should be able to carry forces' moments (torques) in all directions. Hence when trying to predict the 'loadings' on implants, this means, in general, trying to predict the forces and moments on each implant. What makes this problem difficult to solve is the fact that each implant is connected to both the bone and the prosthesis; computing the loads (and stresses and strains) in each part of the structure is a problem that is not solvable by statics alone, but also requires data on the material properties of the implants, bone and prosthesis as well as their stress-strain behaviours.

The main methods for predicting loadings on oral implants consist of two types of analyses.

The first type of analysis is the so-called analytical approach, which is based on using equations taken from conventional engineering textbooks and applied to the case of oral implants supporting a prosthesis. Examples of this approach include the so-called 'see-saw' analysis of loading on two implants by Rangert[10] as well as a more involved analysis first presented in the pioneering 1983 publication of Skalak[8]. The so-called 'Skalak model' idealised the distribution of implants, bone and prosthesis as a special case of a mechanical engineering model used to compute the vertical and horizontal load-sharing among bolts used to fasten together two rigid plates. Skalak, Brunski and Mendelson[9] and Brunski and Hurley[11] then extended this original Skalak model to take account of different axial and bending stiffness values for the various implants in the distribution. Morgan and James[12] did work along the same lines. It is beyond the scope of this paper to present the details of these analytical approaches, but calculations with the Skalak model can be readily done with a spreadsheet such as Excel; indeed, all of the calculations of implant loading per the Skalak model in this paper have been done in this manner.

The second main way to predict loadings on implants is via more sophisticated computational methods, such as finite element analysis (FEA). There are many examples of analyses of implant loading using FEA, e.g. Elias and Brunski, 1991[13]; Ujigawa et al 2007[14]; Naini et al, 2011[15]. The input data in these analyses include the geometry of the bone, implants and prosthesis; the known or estimated material properties of all materials; the boundary conditions between materials; the known or assumed loadings on the prosthesis; and the stress-strain laws for all materials involved. Such models can be relatively straightforward to develop using any number of FEA software packages running on a common laptop, although if the geometry is more complicated – for example when attempting to create a FE model from extensive input datasets derived from computerised tomography (CT) scans – then the computational problem can become large enough to require a more powerful computational platform.

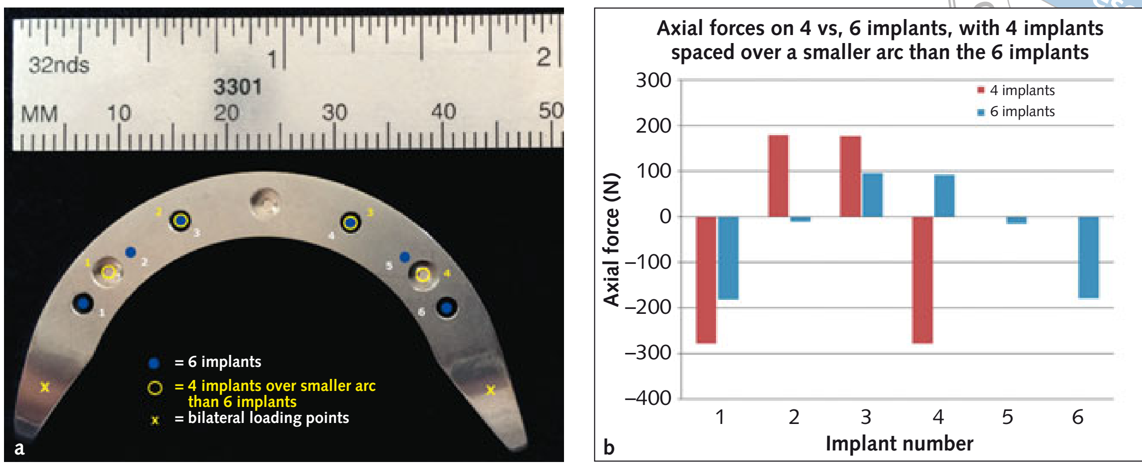
Calculations of implant loading in various situations

Example 1: Is it better to use 4 or 6 implants to support a prosthesis when the 4 implants are spread out over a smaller arc-length than the 6 implants?

Using the Skalak-type analytical model described previously, it is possible to answer this question as follows. Fig 2a shows the labelled undersurface of a prosthetic bar illustrating possible placements of 4 or 6 implants to support a prosthesis; the legend in the image shows where the 4 or 6 implants have been placed, and the yellow X's indicate the locations of the two distal loading points where test forces of 100 N were bilaterally applied. (The anterior of the jaw is toward the top of the figure.) When the 4 implants span a smaller arc than the 6 implants, the 4-implant construction has longer cantilevers than the 6-implant structure – a parameter that definitely influences the loading on the implants.

The results of the axial load calculations with the Skalak-type model (Fig 2b) show clearly that the magnitudes of the axial forces on the 4 implants arranged over the smaller arc is larger than for 6 implants. (By convention in this modelling, a posi-

Fig 2 (a) Image of the undersurface of a titanium prosthetic bar with locations marked for 4 implants spanning a smaller arc-length than 6 implants. (b) Bar graph showing the axial forces on the 4 or 6 implants as computed using the Skalak model.



tive axial force on an implant is tensile, tending to extract it from the bone, while a negative axial force indicates a compressive force on the implant, tending to push it into the bone.) For example, a comparison of axial forces on the distal-most implants 1 and 4 in the 4-implant option (with smaller arc) vs. the distal-most implants 1 and 6 in the 6-implant option shows that the forces are about twice as large for the 4 implant solution. Similarly, for the anterior implants, the force levels are much greater – for example more than twice as large – in the 4-implant solution. It follows from this example that if one's goal is to have smaller axial forces on the implants, then the 6-implant case is 'optimal'.

Example 2: Is it better to use 4 or 6 implants to support a bar when the 4 implants are spread out over the same arc-length as the 6 implants?

This is a similar case to Example 1, except now the 4 implants cover the same arc as the 6 implants. Again the numbered circles in Fig 3a indicate the positions of the 4 or 6 implants, and the distal Xs represent two loading points where test forces of 100 N were applied bilaterally in the comparisons; the legend in the image shows where the 4 vs. 6 implants are placed. When the 4 implants span the same arc as the 6 implants, implants 1 and 4 are at the same distal locations as implants 1 and 6 in the 6-implant distribution. This also means that the 4-implant case has the same distal cantilever lengths as the 6-implant prosthesis.

The results of the axial load calculations with the Skalak-type model (Fig 3b) show that when the 4 implants span the same arc as the 6 implants, the compressive axial loads on the most distal implants 1 and 4 in the 4-implant option are loaded to virtually the same axial force values as the implants 1 and 6 with 6 implants. Likewise, the forces on the anterior implants are similar, with 4 and 6 implants. Notably, two finite element models of essentially this same example – 4 vs. 6 implants spread out over the same arc length – predict the same results as this analysis with the Skalak model[16-18]. From these results it follows that if the goal is to have smaller axial loads on the implants, then there is no significant benefit in selecting 6 rather than 4 implants, as long as the 4 implants span the same arc length as the 6.

Regarding the 4- and 6-implant options discussed in Examples 1 and 2, it is also possible to use the concept of the 'anteroposterior spread (AP spread)' to obtain insight into the pros and cons of various arrangements of implants, although this concept does not provide quantitative information about actual implant loadings; instead it is more of a general guideline for determining a maximum cantilever length. The AP spread has been defined as[19]:

"Distance from a line drawn between the posterior edges of the two most distal implants in an arch and the midpoint of the most anterior implant in the arch. This measurement is used to calculate the maximum posterior cantilever length of the prosthesis, which is usually 1.5 times the AP spread."

Applying the idea of the AP spread to Examples 1 and 2, it is possible to re-examine the merits of

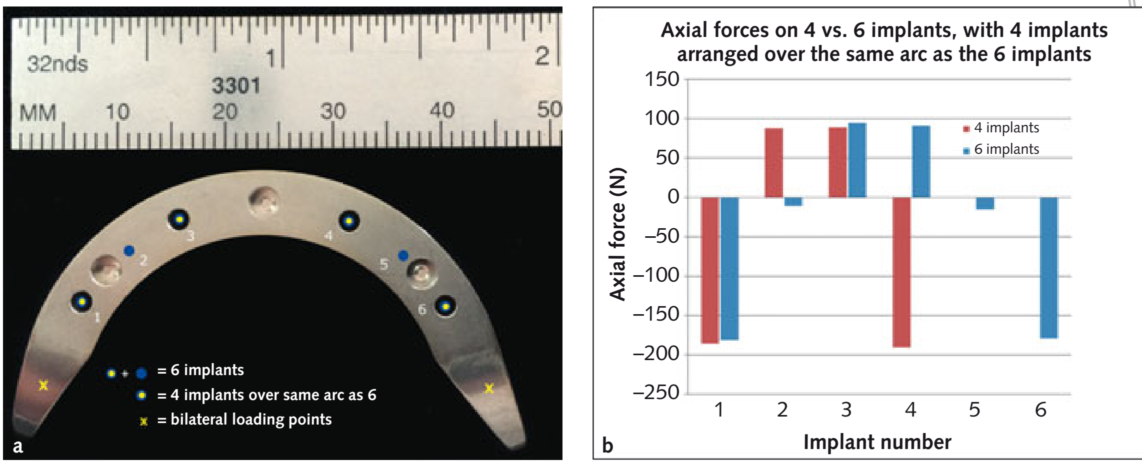


Fig 3 (a) Image of the undersurface of a titanium prosthetic bar with locations marked for 4 implants spanning the same arc-length as with 6 implants. (b) Bar graph showing the axial forces on the 4 or 6 implants as computed using the Skalak model.

4 implants spread over a smaller arc or the same arc as 6 implants. (See also McAlarney and Stavropoulos, 1996[20].) In the former case, the AP spread rule would suggest about 7 mm for the maximum cantilever length, while in the latter case it suggests about 12 mm. Comparing these suggestions to the Skalak calculations in Examples 1 and 2, the Skalak calculations used cantilever lengths of 12.2 mm for the 4 implants over a smaller arc, and 8.6 mm for the 4 implants spanning a larger arc. Comparing these values to what is suggested by the AP spread guideline, this means therefore the cantilever length of 12.2 mm for the 4 implants over a smaller arc is not optimal because 12.2 mm >7 mm. Alternatively, a cantilever length of 8.6 mm for the 4 implants spanning the larger arc (i.e. the same arc as the 6 implants) would be deemed suitable in terms of AP spread, because the cantilever length of 8.6 mm used in the Skalak modelling is less than the maximum cantilever length of 12 mm suggested by the AP spread. So in these examples, the guideline of the AP spread is consistent with the more detailed findings from the Skalak model.

However, it is important to remember that neither the AP spread nor the Skalak model alone is conclusive in defining the optimality of implant loading; ultimately, as discussed later, that issue must also consider the stresses and strains in the interfacial bone, implants and prosthesis, as well as the relationship of those stresses and strains to failure limits for the materials involved.

Example 3: If one uses 3, 4, or 6 implants to support a prosthesis, what differences exist in the loadings per implant, and what is 'optimal'?

A biomechanical comparison of using 3, 4 or 6 implants to support a bar loaded bilaterally by 100 N in the distal locations provides an instructive comparison. As shown in Fig 4a, the legend for labels on the undersurface of the titanium bar describes the placements of 3, 4 or 6 implants. In this example, the positions of the 3 implants are marked and correspond to their locations in the Novum design of Brånemark[21]. The 3 implants span an arc slightly smaller than the 4 and 6 implants in this example, e.g. the cantilever length of the 4- and 6-implant prostheses is a few mm shorter than the cantilever length of the 3-implant prosthesis. The results from the Skalak-type calculations (Fig 4b) show that the axial loads on the implants in the 3-implant distribution are larger than they are for the 4- and 6-implant distribution. In particular, the tensile axial force on anterior implant 2 in the 3-implant treatment option is nearly 300 N, while the maximum tensile force on anterior implants for the 4- and 6-implant options reaches 100 N – a 3-fold difference. The values of the compressive axial forces on the distal implants in the 3-, 4- and 6-implant prostheses are similar, although slightly larger with 3 implants.

However, the above results about forces alone do not tell the whole story vis a vis an evaluation of 'optimality' of 3, 4 or 6 implants; there is more to the analysis. The implants within the 3-implant system

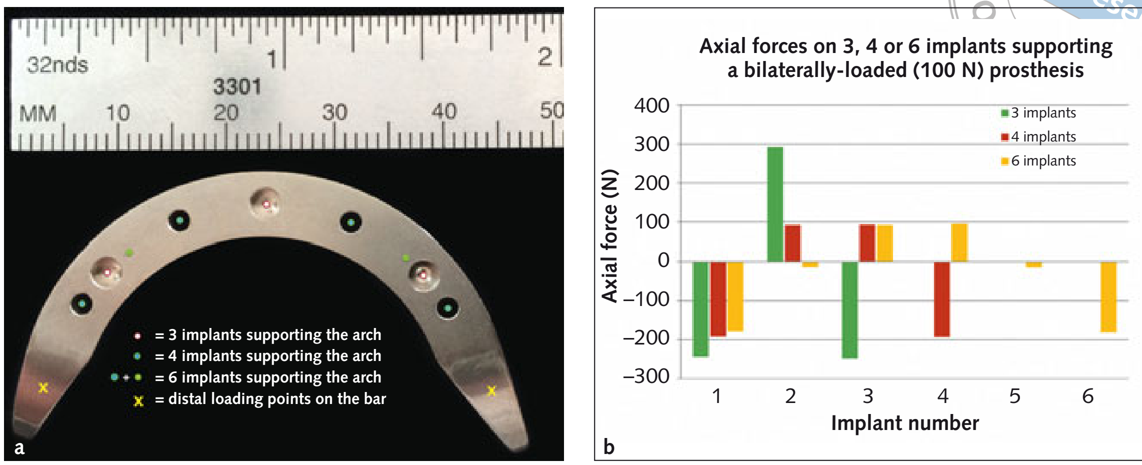


Fig 4 (a) Image of the undersurface of a titanium prosthetic bar with locations marked for 3, 4 or 6 implants; in this example the 3 implants are located as originally planned in the 2001 Novum system of Brånemark, while the 4 and 6 implant arrangements span a slightly larger arc-length than the 3 implants; (b) bar graph showing the axial forces on the 3, 4 or 6 implants as computed using the Skalak model.

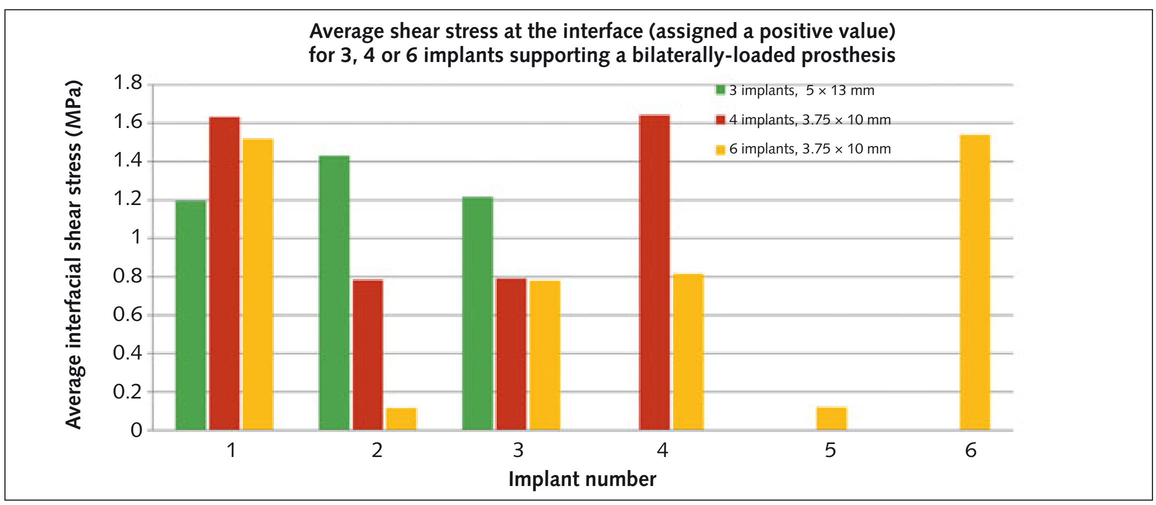


Fig 5 Bar graph showing the average interfacial shear stresses on the 3, 4 or 6 implants analysed in Fig 4. The implants in the 3-implant distribution have larger diameters (and therefore larger interfacial surface area) than the 3.75 mm-diameter implants in the 4- or 6-implant distributions, which explains why the interfacial stresses in the 3-implant case are sometimes smaller than in the 4- and 6-implant situations.

here correspond to the 3 implants used in the original Brånemark Novum system and those implants had a larger diameter (5 mm) than the diameter of 3.75 mm for typical implants used in typical 4 and 6 implant arrangements. As pointed out earlier in this paper, while the axial forces on implants are relevant, so are the resulting stresses in the interfacial bone, and these stresses depend on the implant diameter as well as other factors. So, a critical question is how the stresses in interfacial bone compare in the 3-, 4- and 6-implant options.

An initial answer to this question comes from Fig 5. First consider the average interfacial shear stress for the implants in the 3-, 4- and 6-implant options; these average shear stresses can be estimated by taking the absolute value of the axial force on each implant from the Skalak model and dividing that axial force by the available surface area of each implant. (For the stress calculations, it is not as relevant to be concerned with sign of the axial load on the implants – negative for compression, positive for tension; the key value is the magnitude of the resulting average shear stress.) The approximate surface area of each 5 mm × 13 mm implant is larger than the approximate surface area of each 3.75 × 10 mm implant in the 4- and 6-implant options. So the difference in the data in Figs 4 and 5 is that the forces in Fig 4 have been divided by bone-implant area in order to produce Fig 5. From these stress calculations it is clear that the shear stresses in bone around the two distal implants (1 and 3) in the 3-implant option are actually less than they are in the bone around the two distal implants of the 4- and 6-implant options. At the same time, the average shear stress in bone around the anterior implant of the 3-implant Novum system is about 1.4 MPa, larger than the average shear stress on the anterior implants in the 4- and 6-implant options, although the absolute value of this average shear stress is actually less than the average shear stress on the distal implants in the 4- and 6-implant options. Therefore, if a discussion of 'optimality' of treatment with implants starts to consider the magnitude of the interfacial stresses in the bone (as was recommended earlier in this paper, in Step 3 of the treatment planning analysis), then it becomes clear that there are some benefits of the 3-implant situation with large diameter implants, because the interfacial shear stresses are somewhat lower in the 3-implant option than in the 4- and 6-implant options. This analysis is approximate, because it does not account for details such as screw threads on the implants, amount of bone coverage, properties of the bone, etc., but the gist of the argument remains clear.

Example 4: Is a fixed prosthesis with 5-implants suitable in a maxilla where more than 5 implants were originally planned?

This is an analysis of an actual patient (courtesy of Dr Kenji W. Higuchi, Spokane, WA, USA) where problems in the healing of 2 of the originally-installed 7 implants raised the question of whether the 5 remaining integrated implants would be adequate to support the intended prosthesis in the maxilla (Fig 6). From a biomechanical viewpoint, the question is whether the 5 remaining implants would adequately support loading of the prosthesis, or whether it would make a significant difference if the clinician were to perform a revision surgery to install a 6th implant at a position in the right anterior side (marked by an 'X' in Fig 6), followed by substantial additional healing time (e.g. 5 to 6 months) before a final prosthesis could be considered.

A Skalak model was set up to allow comparison of possible 5- and 6-implant prostheses (Fig 6), for a test load of 100 N being applied over the location of the implant 3 in the images. An inspection of the bar graphs in the two treatment options reveals that there is hardly any difference in the axial forces per implant. Certainly the axial forces are a bit larger with 5 implants, but not significantly larger. Because of this result and additional simulations about the loading (not shown here), the decision was made to go ahead and use the remaining 5 implants to support a Marius denture. The patient had no problems after this stage of treatment.

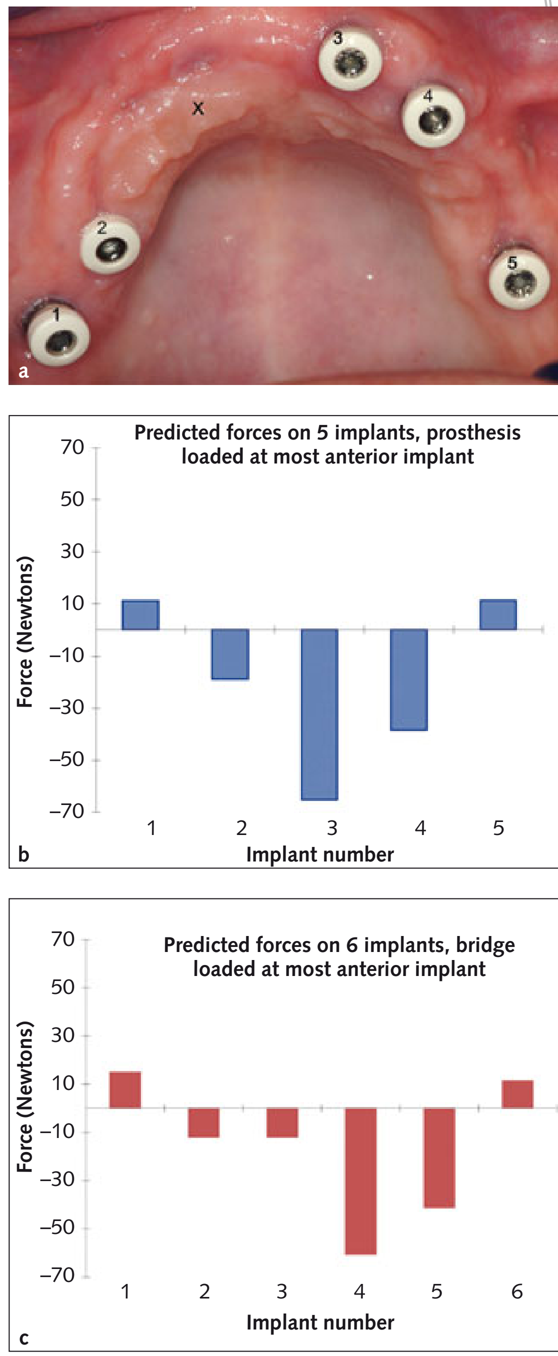


Fig 6 (a) Image of the maxilla of a patient in which 7 implants had been planned, but only 5 implants were properly integrated. (b and c) Calculations of the axial loading per implant (using the Skalak model) revealed little difference in using 5 (b) vs. 6 (c) implants.

Example 5: The biomechanical rationale for tilting an implant: a prelude to the rationale for the 'all-on-4' approach in a full arch

The basic biomechanical aspects related to tilting of oral implants in situations such as the 'all-on-4' approach have been discussed by this author[22] as well as by others[18]. However, before discussing the biomechanical details of tilting in full arch cases and how this relates to 'optimal' numbers of implants, it

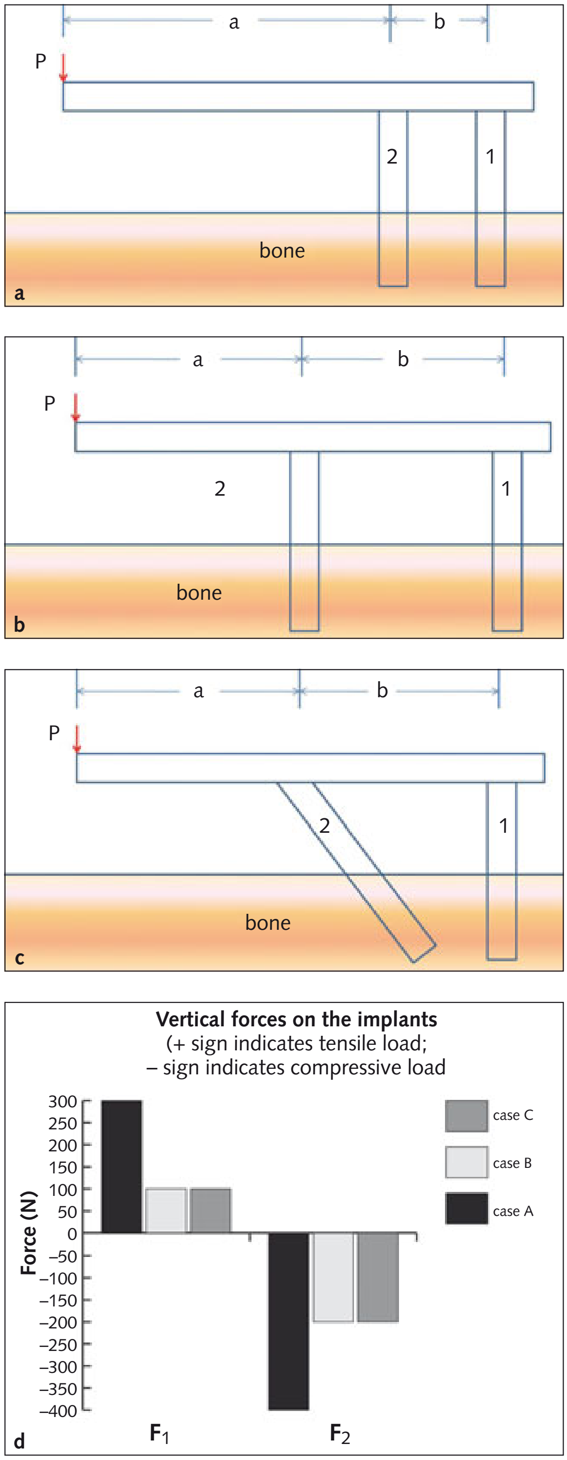


Fig 7 Two-dimensional illustration of the rationale for tilting an implant. The diagrams in (a) to (c) – plus the vertical forces predicted by the Skalak model in (d) – show that if the cantilever distance 'a' is reduced by increasing the implant spacing 'b', the vertical force on each implant can be decreased. The vertical loading on the implants in (b) and (c) are the same because tilting implant no. 2 as shown in (c) produces the same point of connection of the top of that implant to the prosthesis – and the same inter-implant spacing 'b' – as with the upright implant no. 2 in (b).

is first worth analysing the simplest example of the pros and cons of tilting, which can be seen in a 2-implant structure (Fig 7).

For example, in Fig 7a upright implants no. 1 and no. 2 are spaced at inter-implant distance 'b' while supporting a prosthesis loaded by downward vertical force P acting at the end of a cantilever, which is at a distance 'a from implant no. 2. Assuming the prosthesis is attached to the implants by ball-and-socket joints (which means that no moments are supported by the denture-implant junctions), this problem can be analysed using simple 2D statics yielding the following result: the vertical force on implant no. 1, F1, will be tensile (acting vertically upward) with a magnitude equal to (a/b)P; and the vertical force on implant no. 2, F2, will be compressive (acting vertically downward) with a magnitude of (1+a/b) P. Inserting some numerical values into these equations, if a = 30 mm, b = 10 mm, and P = 100 N, then F1 = +300 N and F2 = -400 N (with the + sign indicating a tensile force and the – sign indicating a compressive force). These results are plotted in the bar graph of Fig 7d, along with the results from analysing cases B and C, as follows.

Now if it were possible in a given clinical case to achieve a larger inter-implant spacing (distance 'b' – from, say, 10 mm to 20 mm – the cantilever distance 'a' would then be decreased from 30 mm to 20 mm, which in turn means that the recomputed values of F1 and F2 (using the formulae above) are F1 = +100 N and F2 = -200 N (again with a + sign indicating a tensile force and a – sign indicating a compressive force). The interesting result is that these two vertical forces in case B now are substantially decreased by the increased implant spacing and shorter cantilever, compared to the forces in the situation of Fig 7a.

Given these results, it would be preferable, or 'optimal' – all other things being equal – to arrange two upright implants as in Case B, with the larger spacing 'b' of 20 mm and the smaller cantilever 'a' of 20 mm, because that would give lower forces on the two implants compared to the situation of two implants spaced closer at 10 mm (Case A). However, the key point is that sometimes anatomical factors – such as lack of enough available bone – prevent placing the upright implant no. 2 at the desired larger inter-implant spacing; indeed, this is the anatomical problem originally explained by Krekmanov and coworkers[23].

A benefit of tilting is that it is a way around the problem of lacking enough available bone for an implant where one wants it. The idea is to place the apex of implant no. 2 in available bone stock (perhaps about 10 mm away from implant no. 1, as in Fig 7a) while tilting the top of implant no. 2 so its top now can connect to the prosthesis at the larger, more desirable, inter-implant spacing of b = 20 mm. The see-saw (and Skalak) analysis predicts that this approach will be effective, because the distances 'a' and 'b' in the equations for F1 and F2 are measured at the locations where implants connect to the prosthesis, not the locations where the implants' apices reside in bone. So, for example in Case C (Fig 7c) the tilting of implant no. 2 produces the same downward forces on the two implants as in the upright, 20 mm-spaced implants in Fig 7b, i.e. F1 = +100 N and F2 = -200 N (with the + sign again indicating a tensile force and the – sign indicating a compressive force).

It is also important to realise that although there are identical vertical forces on the implants at the locations where they connect to the prosthesis, there is a major difference between the two situations in Figs 7b and 7c: while calculations predict that the same force F2 acts in a vertically-downward direction at the top of implant no. 2 in Figs 7b and 7c, implant no. 2 is tilted in Fig 7c but upright in Fig 7b. This last fact begs the obvious question: In Figs 7b and 7c, doesn't the tilting of an implant make a major difference in terms of the stresses and strains in the prosthesis, implant and bone? The answer is: "Yes, if the same vertically directed force is acting on the upright and tilted implants, but no if the same force does not act on the upright and tilted implants".

The foregoing can be illustrated with a convenient series of examples in Fig 8 (developed using FE simulations). These simulations illustrate in a simple example that, yes, all things being equal, tilting will cause larger stress and strain in the surrounding bone, and on that basis, tilting might appear detrimental. However, the point about tilting implants is that, in a sense, we are not considering a situation of 'all things being equal'. If we do the tilting effectively, it is possible to decrease the vertical force on the tilted (and other) implants, e.g. compare the forces on the implants in Figs 7a and 7c in the example just discussed. So for instance in Fig 8, when 50 N acts on a tilted implant instead of, say 150 N, then there are smaller tensile and compressive strains in the interfacial bone compared to when 150 N of vertical force acts on either the upright or tilted implant.

The aforementioned is another example of the need to define 'optimality', not just in terms of the

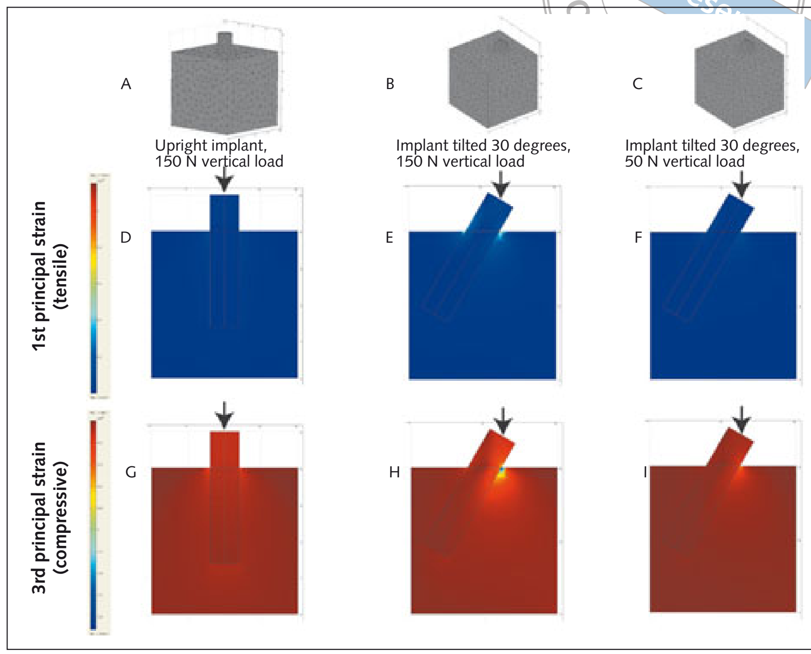


Fig 8 Results from FE simulations of simplified upright (A, D, G) and tilted (B, E, H) cylindrical implants integrated in bone and loaded by the same 150 N vertical force; comparing these two situations, the interfacial principal strains in the bone are larger for the tilted implant. Alternatively if the tilted implant is loaded with a smaller vertical force of 50 N (C, F, I), the interfacial principal strains are not very different from what they were in the case of the upright implant loaded with 150 N. In the images D, E, and F, light blue indicates a higher tensile strain, while in images G, H and I, light-red to greenish-yellow indicates higher compressive strain. (Reproduced with permission from: Brunski JB. "Biomechanical aspects of tilted regular and zygoma implants." Chapter 4, pp. 24–45 in Zygomatic Implants: The Anatomy Guided Approach (Ed. C. Aparicio), Quintessence, 2012[22])

number of implants but also in terms of the stress-strain criteria noted in connection with Step 3 of our treatment planning paradigm (Fig 1). That is, tilting can be safe and effective as long as the overall design of the treatment keeps the implant loading – and the stress-strain magnitudes in the bone – in a permissible range.

Example 6: What is the rationale for an 'all-on-4' approach in a full arch?

It is only a small step from the analysis in Example 5 to the biomechanical rationale of the 'all-on-4' approach, which is that tilting can be a means to effectively increase the inter-implant spacing and decrease the length of cantilevers. This in turn can significantly decrease the vertical forces on the implants as well as the interfacial stresses and strains. This idea is now illustrated with some additional examples of full-arch patient rehabilitations.

For instance, Fig 9a considers two treatment options. The first option shows a bar (the under-

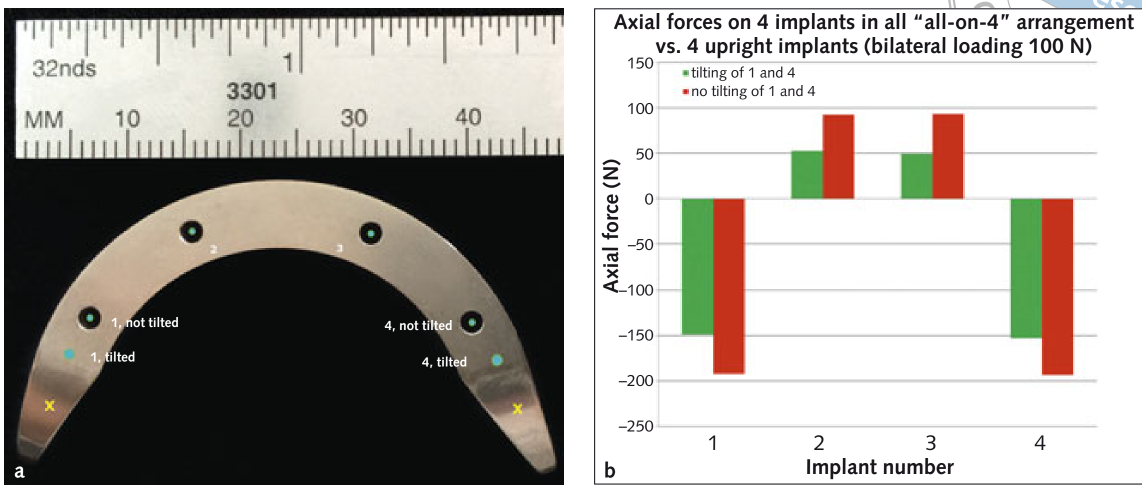


Fig 9 (a) Image of the undersurface of a titanium prosthetic bar showing the locations of 4 upright implants vs. 4 implants, in which the two distal-most implants are tilted. (b) Bar graph showing the vertical forces on the upright vs. tilted implants as computed using the Skalak model.

surface of a Brånemark Novum bar, used in earlier examples) supported by 4 upright implants. The second option shows the same bar supported by the same two anterior upright implants (implants 2 and 3) but now two distally-tilted implants (1 and 4), where the tops of the distal implants 1 and 4 are tilted distally by about 4 mm. (Assuming an abutment height of about 5 mm, this corresponds to a tilting angle of about 38 degrees). The example calculations of implant loading are done with the Skalak model assuming bilateral downward loading of the bar by 100 N at the distal Xs.

The bar graph in Fig 9b shows that in the no-tilting option, the vertical loads on the implants approach -200 N (compression) on distal implants 1 and 4, and about +100 N (tension) on anterior implants 2 and 3. Alternatively, for an 'all-on-4' approach with tilting, this has the effect of decreasing the vertical forces on not only the distal implants 1 and 4, so they are now about -150 N (compression) – but also on the anterior implants 2 and 3 – to about +50 N. Therefore, tilting has substantially lowered the forces on all the implants relative to the non-tilting option, e.g. about a 50% decrease for the anterior implants and a 25% decrease for the distal implants.

Taking this result a step farther, and considering it in terms of the stresses and strains in the interfacial bone (as suggested, again, in Step 3 of our treatment planning algorithm in Fig 1), note in the above example that the tilted implants are not as heavily loaded as their upright counterparts. Now while it is true that a tilted implant exposed to the same vertical loading as an upright implant would typically have larger (possibly less-than-optimal) interfacial stresses and strains, the point is that the tilted implants in the 'all-on-4' structure have less vertical loading than the upright implants located more mesially in our example. Hence, the lower forces diminish concerns about the stress-strain levels in the interfacial bone, the titanium of the tilted implants, and the material of the prosthesis.

To provide a more detailed stress analysis of specific situations involving upright vs. 'all-on-4' treatments, the following examples discuss results from 3-D FE stress analyses of the same prosthesis supported by a) 4 upright implants, or b) 4 identical implants arranged in an 'all-on-4' configuration, in which the two distal implants are tilted (Figs 10a and 10b). The 'upright' and 'all-on-4' options in the FE models are based on the same U-shaped, commercial purity titanium framework (6 mm wide, 4 mm thick) and the same simplified semi-circular idealisation of a mandible of solid bone. In all models, commercial-purity titanium implants (4 × 13 mm cylinders) are assumed to be anchored (bonded) in bone via osseointegration. The distal end of each mandible is constrained from moving in all of the FE models. The distal end of each cantilever of the prosthesis is loaded by a downward force of 100 N. The distal two implants in the 'upright' and 'all-on-4' options have their apices in exactly the same locations; however, in the 'all-on-4' configuration, the top of each distal implant is tilted 30 degrees distally and 10 degrees buccally. The elastic properties of the bone and pure titanium are E = 20 GPa, nu = 0.33 and E = 105 GPa, nu = 0.33, respectively. Also, as a separate exercise, Skalak calculations were used to compute the vertical forces on the upright and 'all-on-4' implants in the two options.

The results from the FE analyses (using Comsol 4.4) of the upright vs. 'all-on-4' options – as well as the results from the Skalak calculations – can be summarised by focusing on 10 selected evaluation criteria that serve as convenient metrics by which to compare the two prosthetic options. As explained in more detail shortly, these 10 criteria include 4 factors characterising stress levels in the prosthesis and implants; 4 factors characterising strain magnitudes in interfacial bone; 1 criterion describing the maximum vertical force on any one implant; and 1 criterion describing the maximum downward deflection of the distal ends of the cantilever sections of the U-shaped prosthetic bar.

The results show that application of the bilateral 100 N loading at the end of the cantilevers elastically bends the prosthesis in each option, creating tensile stresses along the mesiodistal length of each prosthesis; however, these tensile bending stresses were about twice as large in the case of the upright implant configuration, e.g. 79 vs. 44 MPa, respectively (Figs 10c and 10d). Likewise, larger tensile bending stresses occurred on the anterior aspects of the abutment regions of all 4 implants in the upright option compared to implants in the 'all-on-4' option (Figs 10c and 10d). There were also larger compressive stresses in the upright vs. 'all-on-4' option at the locations where the abutment regions of the two distal implants joined the undersurface of the prosthesis (Figs 10e and 10f). Finally, there was a larger downward bending deflection of the cantilever regions of the prosthesis when supported by the upright vs. the 'all-on-4' implants, i.e. 85 vs. 38 microns, respectively; no doubt this result was because of the longer length of the cantilever regions in the upright implant configuration (Figs 10g and 10h). In terms of stress magnitudes that could cause concern about fatigue fracture in titanium, the 10^7 endurance limit for commercial purity titanium is about 300 MPa depending on the exact grade and degree of cold-work of the titanium[24]. Therefore, none of the stress levels developing in the prostheses or implants in the current FE analyses would cause undue concern, although stresses were indeed higher in the upright-implant situation. If loads greater than 100 N were used in the

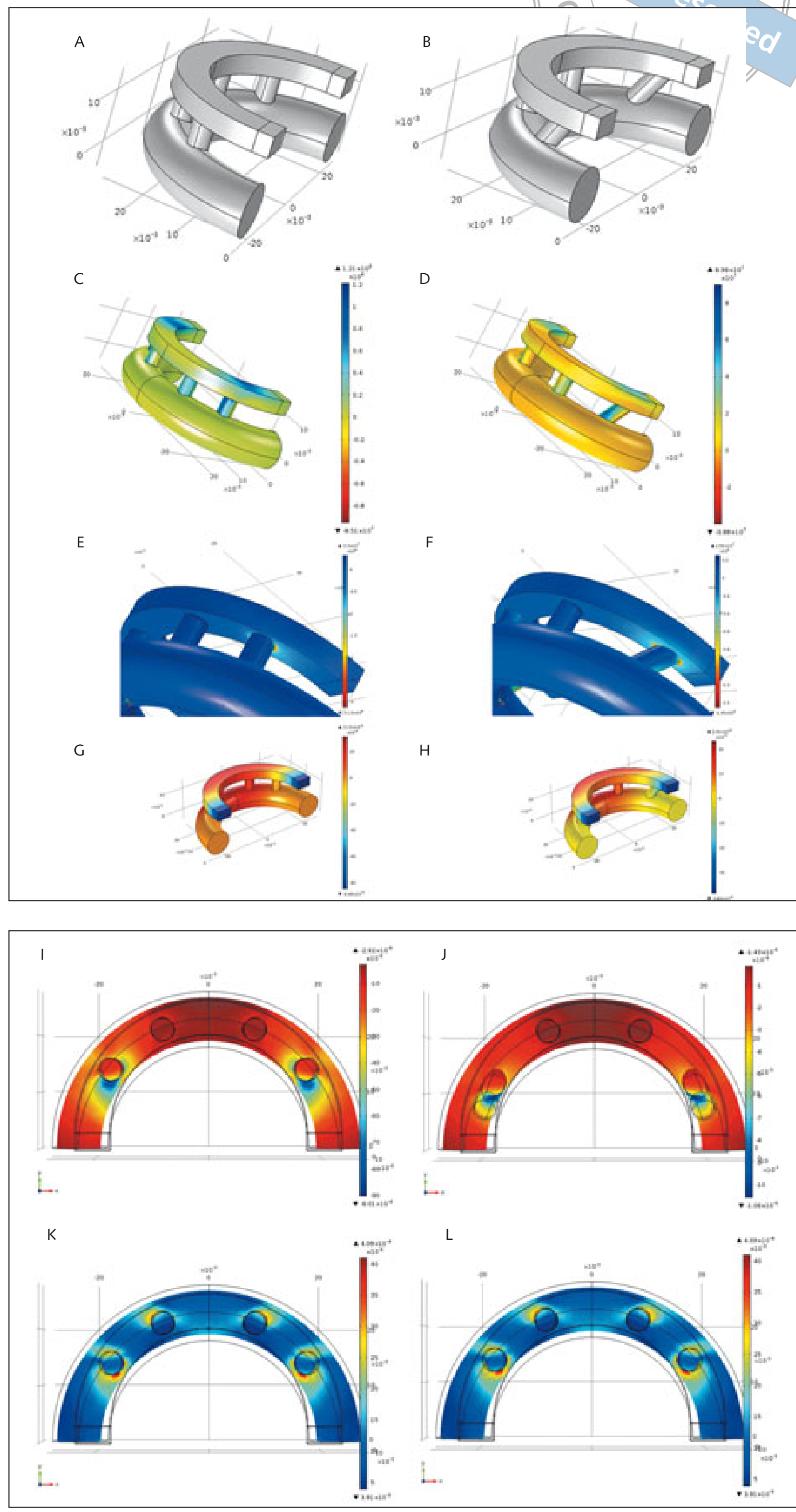


Fig 10 Results from three-dimensional FE analyses of 4 upright implants supporting a bilaterally-loaded bar vs. 'all-on-4' implants supporting the same loaded bar. (a and b) Geometry of the situations. (c and d) 1st principal (tensile) stresses in the upright vs. 'all-on-4' cases. (e and f) 3rd principal (compressive) stresses in the two cases; note junction between abutment and undersurface of bar. (g and h) Vertical (occluso-apical) displacements of the bars in each case; note displacement at the ends of the bars. (I and j) 3rd principal (compressive) strains in bone in a plane of section taken approximately 1 mm below the crest of the mandible. (k and l) 1st principal (tensile) strains in bone in the same section plane as in (i) and (j).

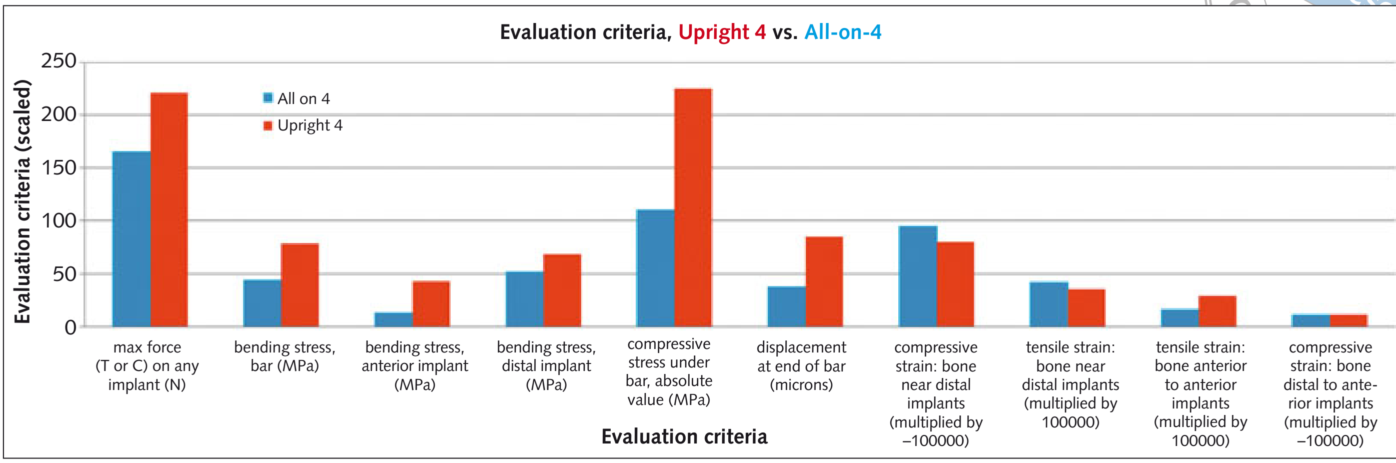


Fig 11 Bar graph collecting 10 criteria for comparing biomechanical conditions found in the 'upright 4' vs. 'all-on-4' simulations discussed in Fig 10 (see text for discussion).

FE simulations, stresses would increase proportionally in both models, so that stresses in the upright-implant option would reach the fatigue endurance limit before the 'all-on-4' option.

Concerning compressive strains in interfacial bone in upright vs. 'all-on-4' options (Figs 10i and 10j, the strain magnitude in regions of crestal bone located distal to the most distal implants was only slightly larger for the 'all-on-4' configuration compared with the upright option, i.e. -0.0945% vs. -0.0805%, respectively, and there was virtually no difference in the compressive strain magnitude at the distal crestal locations around the anterior implants of both the upright and 'all-on-4' configurations. For the tensile strain magnitudes on the distal aspects of the two distal implants in each configuration (Figs 10k and 10l), the strains were somewhat larger for the 'all-on-4' option, i.e. 0.0421% vs. 0.0358% respectively. Also, the tensile strains at crestal locations anterior to the anterior implants were larger for the upright as opposed to the 'all-on-4' option, i.e. 0.0293% vs. 0.0164%, respectively (Figs 10k and 10l). Notably, these magnitudes of strain in bone – peaking at about -0.09% in compression and 0.03% in tension – are below a danger limit of 0.4%, which has been cited as an approximate threshold for fatigue failure in compact bone after about 1000 cycles in tension or 10000 in compression[25]. Therefore, as in the discussion of stresses, none of the strain levels in the bone would cause undue concern in either option – at least for 100 N bilateral loading. (Note that these simplified FE analyses do not account for threads on the implants, which are known to concentrate stress and strain in the bone.) If loads greater than 100 N were used in the FE simulations, strains would increase proportionally in both FE models, and could eventually reach magnitudes that could cause concern.

One last metric of comparison between the upright and 'all-on-4' option is the maximum force occurring on any one implant in the distribution; this maximum force was larger in the upright option than in the 'all-on-4' option, i.e. 221 vs. 165 N.

In reviewing the 10 criteria just discussed, there was a 'tie' in one criterion (compressive strain distal to the anterior implants), but in 8 of the remaining 9 criteria, the 'all-on-4' option had smaller stress magnitudes in the bar and implants, as well as smaller strain magnitudes in the bone (Fig 11). Hence, judging from these biomechanical metrics, the 'all-on-4' configuration ranked better than the 'upright 4 implant option', and could in that sense be considered optimal. These results are also consistent with conclusions from an excellent comparative analysis[18] of 3-, 4- and 5-implant options including an 'all-on-4' option; these authors concluded that: "...the 'All-on-Four' configuration…resulted in a favorable reduction of stresses in the bone, framework, and implants."

Example 7: Is there any benefit in using 'all-on-5' instead of 'all-on-4'?

An answer to this question is evident from Fig 12a, which shows two implant arrangements, the first having the same 'all-on-4' arrangement studied in

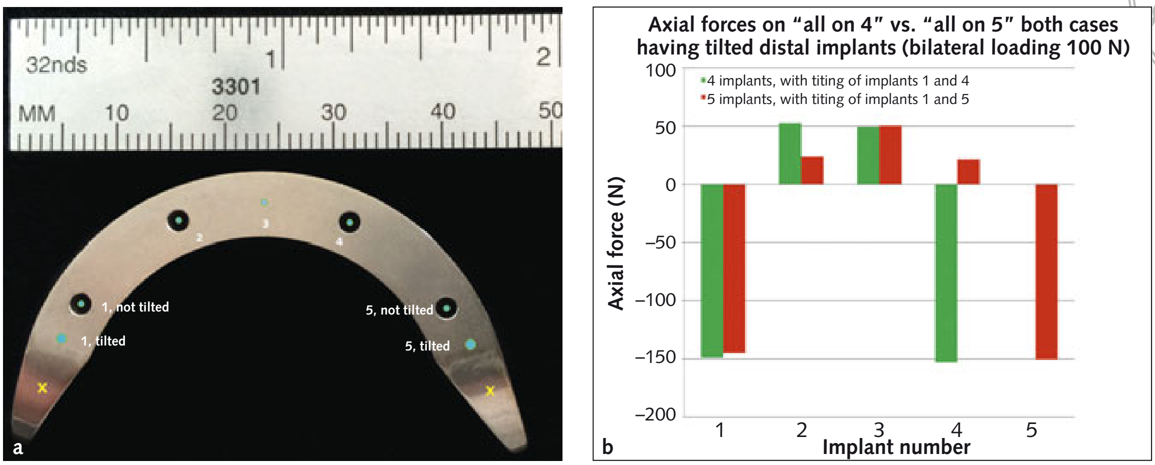


Fig 12 (a) Image of the undersurface of a titanium prosthetic bar showing the locations of 5 upright implants vs. 5 implants in which the two distal-most implants are tilted. (b) Bar graph showing the vertical forces on the upright vs. tilted implants as computed using the Skalak model.

Example 6 (with the two distal implants 1 and 4 tilted) and the second having 5 implants with one 'extra' implant in the middle anterior position – implant 3 – and tilted implants in the 1 and 5 positions that are the same as for implants 1 and 4 in the 'all-on-4' option.

Results from the Skalak analysis of these two situations (Fig 12b) shows that there is little difference between the two cases, i.e. the vertical compressive forces on the distal-most implants are virtually the same in the 4- and 5-implant cases, and so are the tensile loads on the more anterior implants in the two cases. As seen previously, when trying to define the 'optimal' number of implants to use in supporting a full arch prosthesis, biomechanical analyses can help, and in this instance 5 implants in an 'all-on-5' arrangement would be over-designed and inefficient compared to the 'all-on-4'.

Example 8: How accurate are the predictive biomechanical analyses used in this paper? Part 1, in vitro tests

The term accuracy means "…closeness of a measured or computed value to its true value", according to Sokal and Rohlf[26]. Here it is useful to ask whether the vertical forces on implants as predicted by the methods employed in this paper – namely the Skalak model and FE models – are close to the 'true' or actual forces on the implants.

One assessment of accuracy of the Skalak modelling comes from the test results shown in Fig 13[9].

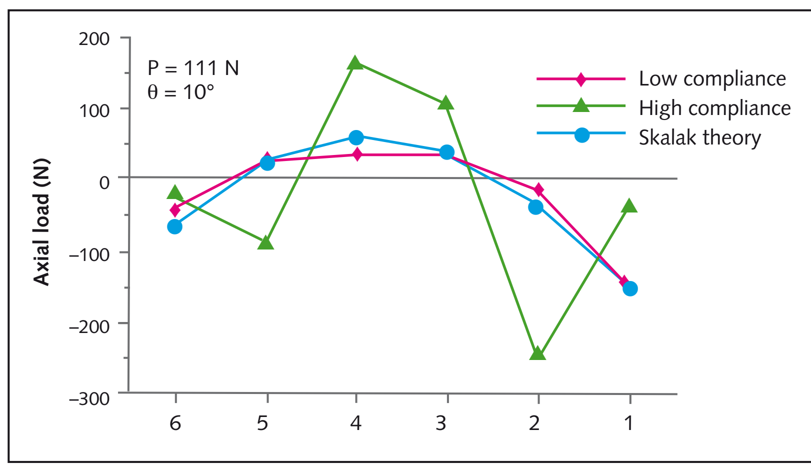


Fig 13 Results from tests of a system consisting of a loaded rigid steel plate ('prosthesis') supported by load-sensing bolts ('implants') whose axial stiffness values could be adjusted[9]. The plate is loaded by 111 N at a cantilever position distal to implant no. 1. Results from Skalak calculations are not shown for the 'High compliance' case; see reference [9] for this detail.

In this testing, a laboratory setup was devised so that the experimental conditions were as close as practical to the assumptions inherent in the Skalak model, i.e. spring-like bolts connecting infinitely rigid plates. To that end the experimental model consisted of strain-gauged load-sensing steel bolts joining two rigid steel plates. (The bolts were analogous to implants while the plates were analogous to the jaw and the prosthesis. The strain-gauged bolts were also known to provide accurate experimental measurements of the axial loading.) The top plate was loaded with vertical forces in different locations, while the vertical forces were then measured using the strain-gauged bolts. The aim of the test was to compare the Skalak model predictions to accurate

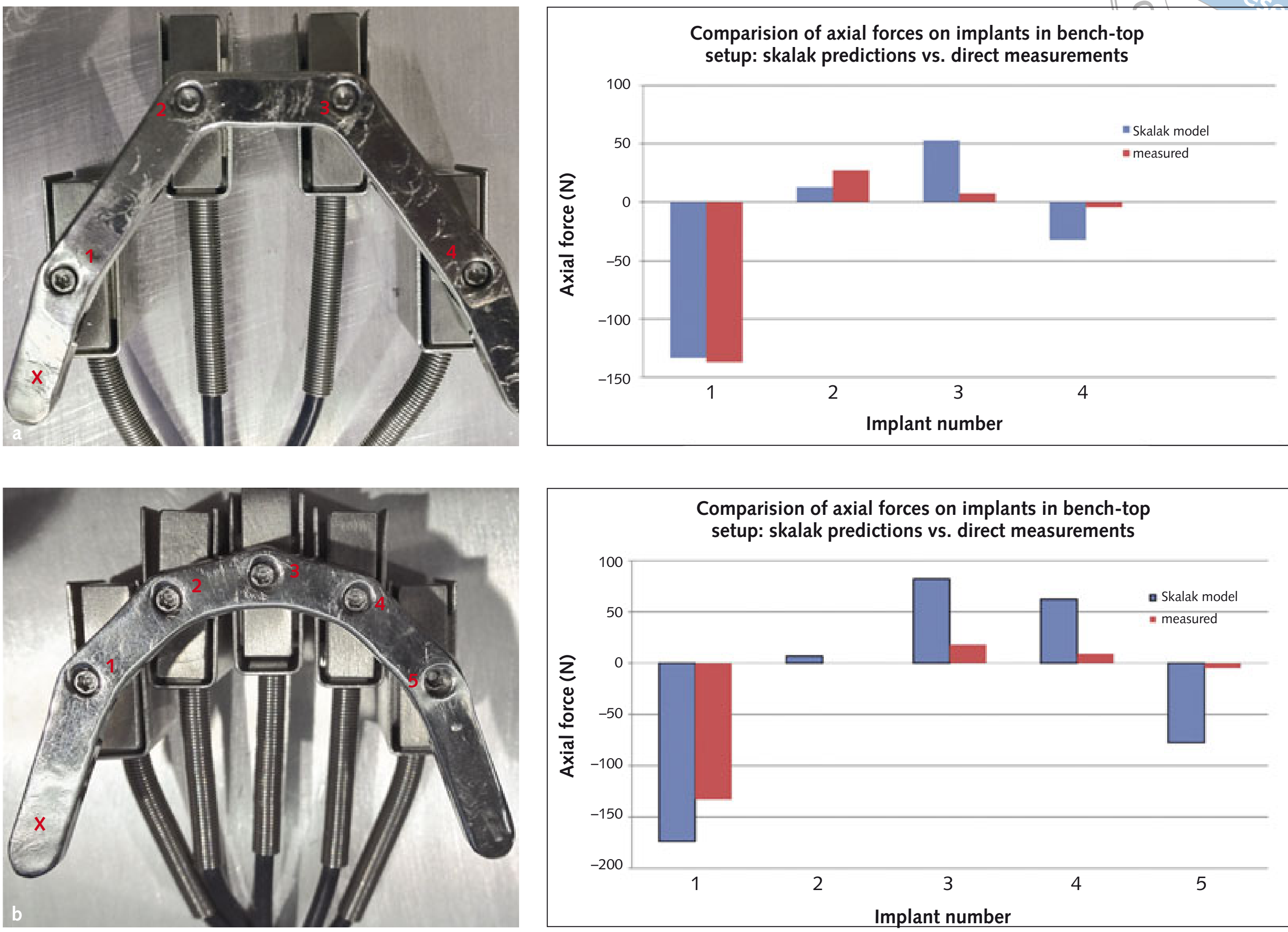


Fig 14 Comparison of experimentally-measured vs. predicted vertical forces on 4 (a) or 5 (b) implants supporting titanium bars loaded at the location of the 'X' in each figure. Predicted forces came from the Skalak model (Bench-top loading system designed and built by Mr Steve Hurson, Nobel Biocare USA).

measurements of the axial loads on each bolt (implant). Fig 13 illustrates that the agreement between the Skalak predictions and measurements was excellent – both when all bolts (implants) had the same axial stiffness and also when the stiffness values of two of the bolts (implants) were decreased. So in this experimental system, it was evident that the Skalak model had reasonably high accuracy in predicting axial loading on multiple bolts (implants) supporting a rigid plate.

In a similar manner, experiments were then conducted to make comparisons of Skalak model predictions vs. measurements of implant loading using a bench-top system designed by Mr Steve Hurson of Nobel Biocare in Yorba Linda, CA, USA (Fig 14). This system consisted of titanium prostheses supported by either 4 or 5 implants connected to 4 or 5 separate force transducers mounted beneath the implants; the force transducers were able to measure the vertical forces on each implant when the prosthesis was loaded at any point using a loading device (not shown). For analysis of the 4 and 5 implant cases with the Skalak model, we measured the (x, y) spatial coordinates of implant locations and points on the prostheses where vertically-downward test loads were applied near the end of the left-hand side cantilever (red X in Fig 14).

The force analysis (Fig 14) allowed comparisons of the measured vertical (axial) force on each implant (red bars) with the forces predicted on each implant via the Skalak model (blue bars). In both the 4- and 5-implant cases, the Skalak model reasonably accurately predicted the vertical force on the implant nearest to the applied loading (implant 1), both in sign (compressive) and in magnitude. For the vertical forces on the rest of the implants, the Skalak model was reasonably accurate in predicting the signs of the forces – including the tensile forces on the anterior implants (2, 3 and 4 in the 5-implant case, plus 2 and 3 in the 4-implant case) as well as the compressive forces (on implant 5 in the 5-implant case and implant 4 in the 4-implant case) – but was not accurate in predicting the true values of the vertical forces on these other implants.

In these bench-top laboratory experiments, the likely reason for the imperfect agreement between the Skalak modelling and the measured forces has to do with the deformability of the prosthesis[13]. That is, the underlying theory of the Skalak model assumes that the prosthesis and jaw are idealised, rigid structures that do not deform under loading, but of course it is known that real materials and structures, including typical full-arch dental prostheses, are deformable, e.g. prostheses do deform even if they are made of metallic or acrylic materials that appear to be ‘rigid’ to the naked eye. Experiments and FE modelling of metal-backed and all-acrylic prostheses (Fig 15) confirm that as the prosthesis becomes more deformable (less rigid), the implants nearest the loading point on the prosthesis take a larger share of the applied load – which, in turn, causes less sharing of loads among all the implants in the distribution. For example, in Fig 15, when the prosthesis is loaded at the cantilever near implant 1, the Skalak model (which assumes an infinitely-rigid prosthesis) under-predicts the forces on implants 1 and 2 and over-predicts the force magnitudes on implants 3, 4, 5, and 6. However, a more accurate FE simulation of the implants and prostheses – which takes into account prosthesis deformability – shows closer agreement between predicted and measured forces.

The role of deformability of the prosthesis in load-sharing among implants was also evident in the results of FE models of 4 vs. 6 implants supporting a titanium prosthesis[16,17]. These workers developed a FE model in which 4 or 6 implants were evenly spaced along the 47 mm of arc between the mental foramina. The implants were attached to a titanium prosthesis loaded with a 100 N vertically-downward force plus a 10 N lingually-directed horizontal force that were both applied along the cantilever region

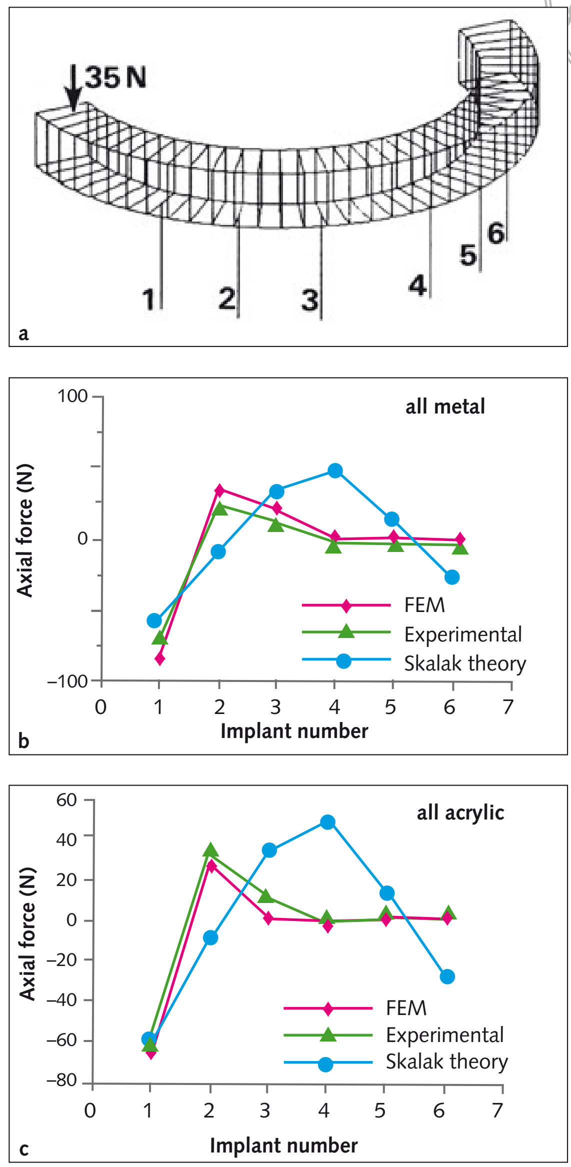


Fig 15 Comparison of experimentally-measured vs. predicted vertical forces on 6 implants supporting an all-acrylic or all-metal U-shaped prosthesis. Here forces were measured using strain-gauged abutments, while forces were predicted using FE modelling and the Skalak model, as indicated in the lower two plots. Clearly the structural rigidity of the prosthesis, which depends on modulus and cross-sectional dimensions, affected load-sharing among the 6 implants. (For more details see Elias and Brunski, 1991[13].)

– which was 8 mm or 16 mm long – on the left side of the mandible. Data on the axial forces on each implant in the 4 vs. 6 arrangement – for both 8 mm vs. 16 mm cantilever lengths – are shown in Figs 16a and 16b (which are based on the present author's plotting of tabulated data in the 1991 paper of Mailath et al[16]). Also plotted in Figs 16a and 16b are results from Skalak modelling of the same cases. Two interesting findings from these data are: a) the 4 and 6 implant arrangements over the same arc show virtually the same axial forces on the 4 or 6 implants – as has already been discussed in earlier examples in this article – and this is true for both the FE and Skalak modelling; b) the values of the axial loads on the implants as predicted by the FE modelling

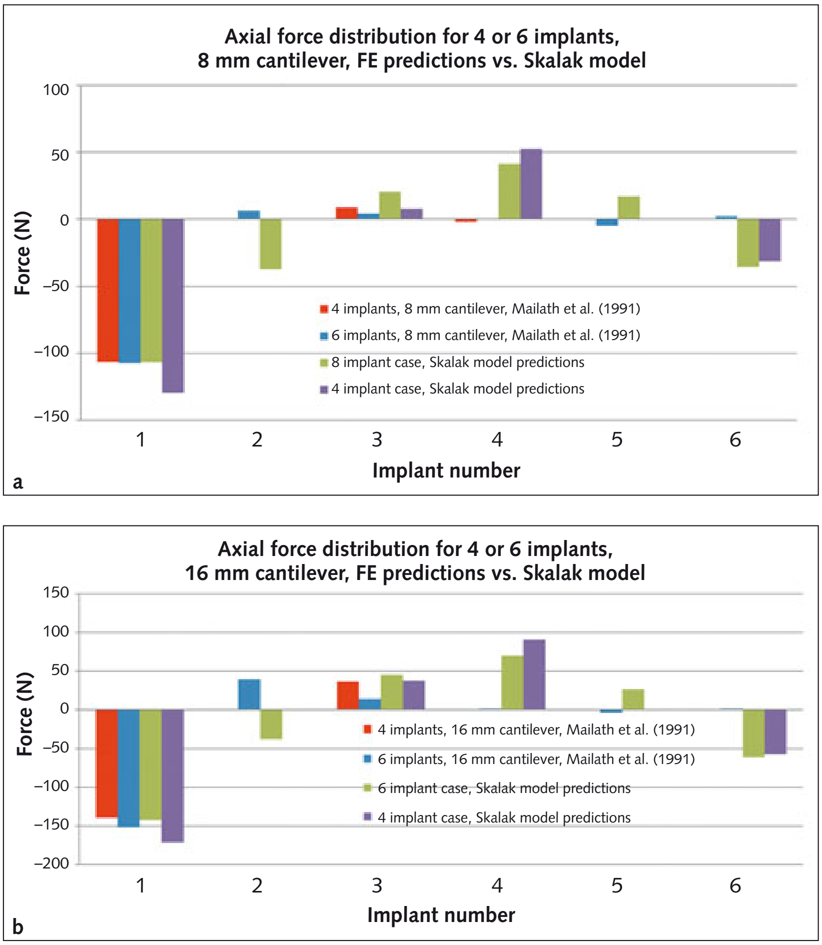


Fig 16 Vertical forces on 4 or 6 implants supporting a deformable prosthesis with a 8 mm (a) vs. 16 mm (b) cantilever, as predicted using FE methods[16] and the Skalak model. The same trend as seen in Fig 15 is seen here: a more deformable prosthesis does not allow as much load sharing among implants as would be predicted by the Skalak model, which assumes an infinitely rigid prosthesis.

do not agree, quantitatively, with predictions from the Skalak modelling, although there is reasonable qualitative agreement between the FE and Skalak modelling. On the last point, the FE model of Mailath et al[16] accounts for deformability of the prosthesis, whereas the Skalak model does not – the same finding that was discussed in the preceding paragraph.

Example 9: How accurate are the predictive biomechanical analyses used in this paper? Part 2, *in vivo* tests

To assess the accuracy of force predictions in actual *in vivo* studies with implants, one approach is to compare the accurately-measured vertical forces on oral implants in humans with predictions of the vertical forces on implants as computed using the Skalak model. To this end, data were available from two male patients (based on data gathered using load-sensing abutments[27]. Each patient had 6 implants supporting a full-arch prosthesis. In the two patients to be discussed below, Case 'H' involved implant-supported prostheses in both jaws, with force data taken only from the mandible, while Case 'C' involved maxillary implants opposed by a natural dentition. Special metal prostheses were used in the patients when measuring the forces on the implants, because these prostheses had special markings allowing the patient to bite down on a special bite fork placed at specific, known locations around the arc of the prosthesis. Before the metal prosthesis was placed, the original Brånemark-style abutments were removed and replaced by special load-sensing (strain-gauged) abutments of 5.5 mm height, which also fit passively with the denture. After the denture was installed over the load-sensing abutments, the patient was asked to bite on a bite fork to measure the biting force exerted at specific locations on the prosthesis, e.g. two distal locations and one anterior location, while the data on the vertical forces on all six implants was collected following the methods outlined in Duyck et al[27].

Meanwhile, to predict the vertical forces on the same set of implants at each loading event, the (x, y) coordinates of each abutment as well as the locations of the applied biting force (50 N) on the prosthesis were input into the Skalak model[28]. In the results presented here, data are discussed for the case of a 50 N bite force exerted at three locations on the prosthesis.

Results from Cases H and C (Fig 17) reveal trends resembling those seen in the *in vitro* tests discussed previously in Example 8. That is, the Skalak model under-estimated the vertical forces on the implants for each of the three loading points with the 50 N force on the prosthesis. For instance, *in vivo*, when the 50 N applied load acted on the prosthesis near implant 1, implants 1 and 2 sustained more vertical force than predicted by the Skalak model. Likewise, the Skalak model under-predicted the loading *in vivo* for the anterior implants when the biting force acted on the prosthesis in the anterior region. The reason for this discrepancy is most likely the same as in the *in vitro* trials of Example 8, i.e. the actual deformability of the prosthesis vs. the assumed infinite rigidity of the prosthesis in the Skalak model. Evidently,

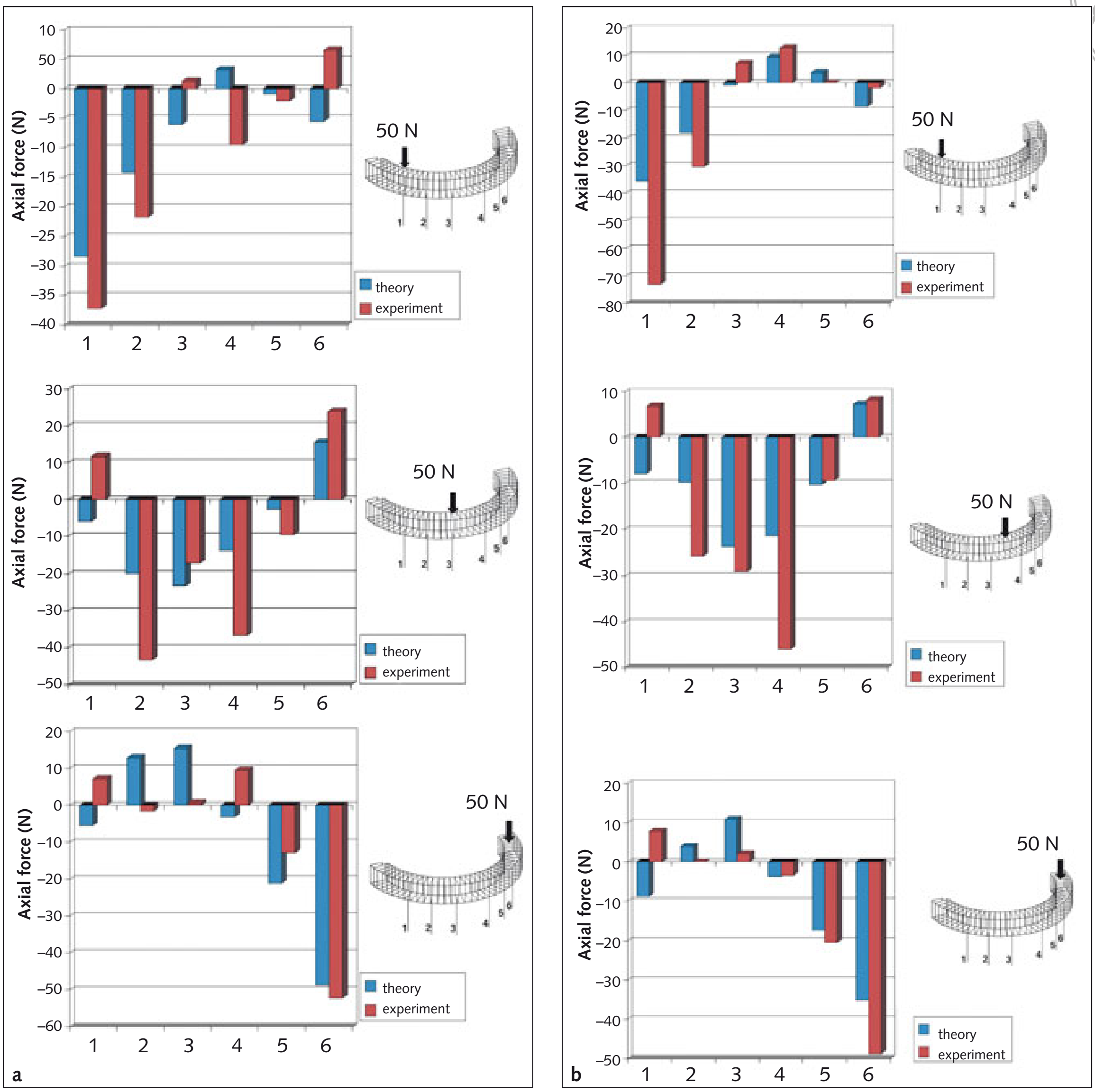


Fig 17 Comparison of measured vs. predicted (Skalak model) vertical forces on implants *in vivo*, based on data taken from work of Duyck et al, 2000[27]. Cases 'C' (a) and 'H' (b) are results from two different patients.

the Skalak model's numerical predictions are not of high accuracy when compared to actual *in vivo* data, although if one considers the model's qualitative ability to predict trends in implant loading, then the overall accuracy is sufficient to allow this model to serve as an approximate guideline in treatment planning.

Besides prosthesis rigidity, two other factors can significantly influence the accuracy of predictions with the Skalak model. The first factor relates to bone-implant stiffness, which is assumed to be the same for all implants in the simplest version of the Skalak model, but which can be varied in the more sophisticated version of the Skalak model[9]. For example, if one had data on the stiffness of each implant in the human trials performed by Duyck et al[27] (Fig 17), then it would have been possible to incorporate that data in the Skalak model to see if there would have been better agreement between experimental and predicted values of forces.

The second source of mismatch between the measured forces and forces predicted by the Skalak model is deformability of the mandible[29,30]. It is known from previous work in human patients[31] that when a patient simply opens the mouth wide while wearing a metal prosthesis attached to load-sensing abutments (Fig 18a), forces and bending moments develop on the abutments (Figs 18b and 18c). In this instance, the magnitudes of the forces and moments are at the low end of the range of typical forces and moments measured during chewing or biting, e.g. a few N and perhaps 10 N-cm, respectively[6]. Notably, such loadings occur simply as

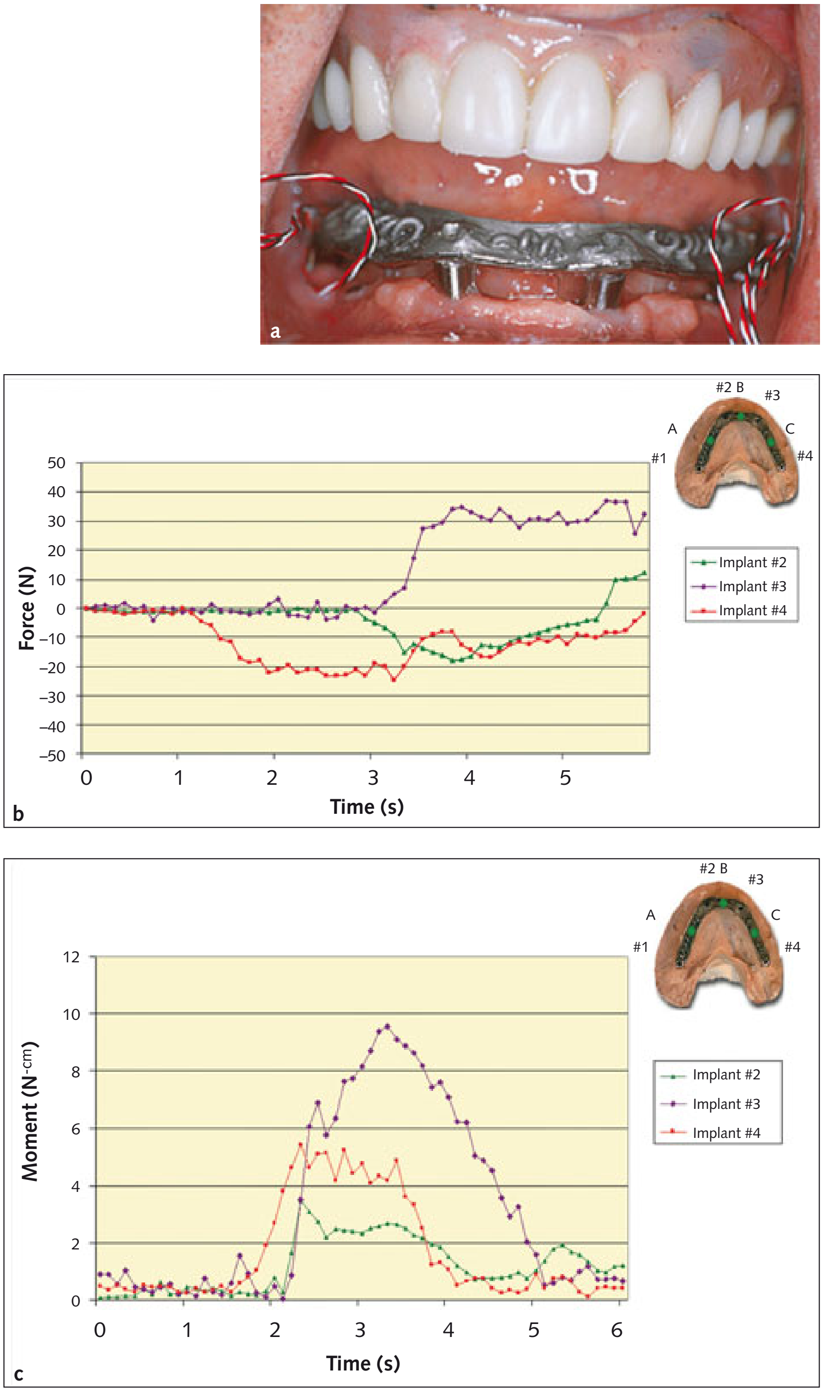


Fig 18 Example of *in vivo* forces and moments on implant abutments supporting a metal prosthesis in a patient who is asked to open his jaw at about the 1-s mark in the plots; vertical force components occur (tensile and compressive, depending on the implant) as well as bending moments[31,34].

a consequence of jaw opening, without any biting or chewing directly on the prosthesis. The explanation for this finding is likely due to human mandibular deformability plus the metal prosthesis rather rigidly attached to the mandible via the implants. It is likely that the intraoral situation becomes analogous to a standard bone plate screwed to bone to prevent or limit motion at a healing fracture site. In other words, when a bone plate is attached to a bone to stabilise a fracture site, the plate carries some of the loading that occurs on the bone; this is accomplished by making sure that the structural stiffness of the plate (and its firm attachment to the bone using screws) produces a stiff system that can support greater or lesser degrees of the loading of the bone, depending on the relative stiffness of the healing fracture vs. plate[32,33]. However, when using the Skalak model to predict implant loading, there is no allowance in the model for jaw flexion; the model assumes that both the prosthesis and jaw are infinitely rigid (undeformable). Therefore, the Skalak model will not predict implant loading from simply opening the jaw – a likely source of the numerical disagreement between the Skalak model predictions and actual measurements taken in the study by Duyck et al[27].

Conclusions

The optimal number of implants to support a full arch prosthesis is predicated on a biomechanical definition of this term; 'optimal' must be broad enough to go beyond just describing the number of implants, and also needs to consider where the implants are placed in the jaw, what sort of bone they are anchored in, what magnitudes of stress and strain develop in the bone, implants and prosthesis; and the relationship of the stresses and strains to thresholds for damage to bone and prosthetic parts. In general, a complete biomechanical treatment-planning regimen should include attention to all of these subjects.

In order to integrate more biomechanical approaches with clinical treatment planning, there are existing aids that can help a clinician predict implant loading. Examples of methods include the Skalak model as well as more involved finite element modelling. While the Skalak model is not always perfectly accurate when used to predict *in vivo* loadings, it can nevertheless provide a reasonable initial analysis of the biomechanical circumstances surrounding a proposed treatment. Increasingly, user-friendly finite element methods can also assist treatment planning, although using such software does require an engineering background in order to use it effectively.

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