CHAPTER 7

Structural analysis for pre-surgery planning:
two applications in dentistry and urology

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Abstract

The chapter focuses on current problems and developments in two areas where
biomechanics, and numerical structural analysis in particular, could be useful:
designing dental prostheses and implants, and conducting structural evaluations
of the pelvic floor.

For the first area, different types of dental implants and prosthetic designs used
for partially or totally edentulous patients were considered in order to analyze the
strain-stress state in bone, which determines whether the biomechanical system
will succeed or fail.

In the second area of investigation, our aim was to demonstrate the impor-
tance of restoring pelvic organ supporting structures (the urethro-pelvic and pubo-
vesical ligaments) following resection surgery performed for uterine or bladder
neoplasias, etc.

1 Introduction

Biomechanics is a part of bioengineering that can be defined as the application of
mechanical engineering concepts and methods to the investigation and solution of
problems in medicine and biology.

While the scope of biomechanics is straightforward enough, the ways and means
it uses to pursue its aims vary widely.

Though the field abounds with abstrusely theoretical basic research, which we
respect since not to do so would be to set limits on intellectual enquiry, there are
very few practical applications. Apart from these theoretical efforts, certain investigations call for some form of clinical follow-up, though without working out the details. What little has been done to solve clinical problems has invariably sprung from interdisciplinary work, and goes hand in hand with methodological advances.

We elected to study biomechanics because of its ability to support research and clinical applications. Thus seen, biomechanics can make an enormous contribution to surgical practice.

The turning point in establishing engineering’s potential role in biological and clinical studies came towards the end of the 1960’s, when it was realized that the theoretical and experimental methods of mechanical engineering could be focused on biological structures in general, and on bone in particular [1, 2].

Thus, a biological structure could potentially be treated as a normal engineering material. This is particularly true of its ability to withstand stresses, including fatigue stresses, and its structural behavior. To handle biological structures in this way, it was necessary to be able to calculate the forces acting throughout the system, to characterize materials, to perform structural analysis and, consequently, to introduce numerical techniques such as the finite element method [3].

Investigations of the “human system” and the prosthetization of any of its parts must thus start from an awareness that any form of surgical intervention will invariably affect the strain-stress state of the structures involved, either directly or indirectly.

Even with the approximations that most numerical models introduce for the mechanical properties of biological tissues (e.g., their isotropy, homogeneity and linear behavior), for the conditions at the interface with prosthetic devices, and for loading and constraint conditions, structural analysis has made significant progress since 1960. For orthopedic implants in particular, structural analysis has made it possible to formulate general rules that now enable us to understand which approaches should be ruled out before proceeding with actual prosthesis design.

The experience gained in one of the most mature areas of structural biomechanics, viz., orthopedic biomechanics, encourages us to apply the methods of numerical structural analysis to assessing the behavior of other biological structures that undergo structural changes following surgery designed to restore a failed function or to remove, for example, neoplastic formations.

2 Dental biomechanics

2.1 Question: what influence do different prosthesis configurations have on bone stress/strain pattern?

The implant-supported prosthesis is an alternative to conventional removable dentistry: while conventional dentures may meet the needs of many patients, others require more retention, stability, function and aesthetics. A review of recent literature [4] indicated that implants used to support an overdenture have a very high success rate, and are thus likely to become more and more widespread.
Today, similar clinical situations can be handled using a variety of prosthetic solutions: in particular, the implant support can differ according to the type of implants used and their layout. Clinical comparisons of different surgical treatments are difficult, as each patient’s biomechanical situation is unique, and the scientific literature provides no clear guidance to the alleged benefits claimed for specific types of dental implant and their morphological characteristics [5].

It is known, however, that the success or the failure of implants interfaced with bone (e.g., orthopedic and dental implants) depends on the structural condition of the biomechanical system made up of the bone structure and the implant [1, 6], providing that the biological reaction is favorable. A good understanding of the strain/stress pattern makes it possible to establish whether bone maintenance, resorption or addition is more likely to take place [7]. Hoshaw et al. [8] applied a dynamic axial tensile load for 500 cycles per day for five consecutive days to Brånemark implants inserted in rabbit tibiae. The result was bone loss around the implant neck, a region where finite element analysis showed high strains. Duyck et al. [9] found crater-like bone defects as a result of a dynamic transverse load applied on Brånemark implants inserted bicortically in rabbit tibiae. The interpretation was that the bone loss had been caused by excessive stresses. Roberts et al. [10] reported a high remodeling rate around the tops of implant threads.

All of these studies confirm that analyzing stress patterns can provide useful guidance in selecting the type of implant to be used.

Analyzing a biomechanical system of this kind is complicated because of the different structures involved (compact bone, cancellous bone, gum, implant, prosthesis), which feature complex geometries and dissimilar mechanical properties. This makes load transmission from the teeth to the bone difficult to evaluate intuitively. As a result, the finite element method (FEM) is needed in order to perform comparative evaluations whereby different surgical approaches to the same bone situation can be simulated.

### 2.2 Analyzed prosthesis configurations and the findings of structural analysis

By way of example, we will illustrate two cases of partial or full edentulism that can be treated with different prosthetic solutions.

#### 2.2.1 Case I

Two different kinds of implant supports for overdenture retention were compared. The two types differed in number of implants, size, location in the mandible and in whether or not a bar is used to connect the implants.

The first solution (fig. 1(a)) for overdenture retention will be referred to below as the ‘conventional’ design, and simulates the insertion of two mutually parallel Brånemark implants in the chin area. An acrylic saddle serves as the base for the prosthesis and is attached to these implants.

The second solution (fig. 1(b)) will be designated as the ‘modified’ design, and simulates the insertion of four screw-type rootform implants anchored to the chin area with bicortical fixation. These implants differ in orientation, and are connected...
Figure 1: FEM models of (a) mandible with a ‘conventional’ implant support design, (b) mandible with a ‘modified’ implant support design.

Figure 2: X-ray image (left) and picture (right) of the ‘modified’ implant support design.

Table 1: Mechanical properties of materials.

<table>
<thead>
<tr>
<th>Materials</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>13,000</td>
<td>0.3</td>
</tr>
<tr>
<td>Trabecular bone</td>
<td>300</td>
<td>0.3</td>
</tr>
<tr>
<td>Gum</td>
<td>20</td>
<td>0.3</td>
</tr>
<tr>
<td>Titanium</td>
<td>100,000</td>
<td>0.3</td>
</tr>
<tr>
<td>Resin</td>
<td>2000</td>
<td>0.4</td>
</tr>
<tr>
<td>Damping layer</td>
<td>500</td>
<td>0.4</td>
</tr>
</tbody>
</table>

to each other by a metal wire by means of a syncrystallization process (fig. 2) [11]. The acrylic saddle used as a base for the prosthesis is attached to this metal wire. A plastic layer is placed between the wire and the saddle to perform a damping function.

A fully osseointegrated condition was simulated (secondary stability). The numerical models consisted of approximately 33,000 4-node tetrahedron elements. Modeled materials are listed in table 1, while mechanical properties agree with data found in the literature [12, 13].

The two models were asymmetrically loaded at the second premolar.Loads were distributed, simulating occlusal contact with the corresponding upper tooth.
The vertical component of the load was 50 N, while the distal-mesial component was 50 N [9]. Constraints simulate muscle action during mastication.

Analysis focused on identifying the cortical and trabecular bone stress/strain pattern and thus determining whether the structural condition was favorable to bone remodeling. von Mises stresses were first considered in order to locate the most highly stressed areas. A more detailed analysis was then carried out to assess the major direction of stress in these areas.

The analysis of von Mises stresses in cortical bone indicated that the peak stress occurs at the implant nearest to the applied load (figs 3(a), (b) and 4(a), (b) in both implant support designs. Peak stress was located on the distal side at the point of implant insertion into cortical bone.
A more detailed stress analysis indicated that the peak stress is due to a notch effect: the stress field is typically triaxial, while the stressed area is very small and matches the clinical evidence of conical resorption [8]. The analysis also showed that the most influential force component is exerted along the $y$ (distal-mesial) direction, given that the force application point was nearly aligned with one of the constraint points, along the $z$ (vertical) direction.

On the whole, the ‘modified’ overdenture produced smaller stresses than the ‘conventional’ design, as von Mises stresses were 34% lower.
This can be explained by the fact that load is distributed over a larger number of implants. In addition, the notch effect is reduced whenever there is more than one discontinuity: stress is more evenly distributed, even if the average stress level rises.

The numerical results are corroborated by clinical experience [14] and by radiographic images showing larger alveolar bone losses (typical resorbed cones) at the point of Brånemark implant insertion in the bone. These observations are borne out elsewhere in the literature [15].

A more detailed analysis was performed in order to assess the structural importance of the metal wire connecting all implants in the ‘modified’ solution. A hypothetical model without wire was developed for this purpose. The numerical analysis demonstrated that eliminating a 2 mm diameter wire produces a 5% higher peak stress. The reduction in implant-bone system stiffness was moderate: assuming that secondary stability has been achieved, the implants are linked to each other by means of cortical bone, which has a lower Young’s modulus than the metal wire but provides a much better geometry because of its larger size.

The results would have differed had we considered primary stability: as the implant would not yet be osseointegrated, it would be constrained in the cortical bone by contact forces alone, thus making the wire’s role in restricting implant bending more important.

Analysis of the stresses in the trabecular bone indicated that the most highly stressed area for the ‘conventional’ implant was located at the distal tip of the implant opposite the loaded area. These stresses can be disregarded for two main reasons: their magnitude was low [16], and their location was far from the proximal implant area, which is the most critical as regards bone remodeling.

For the ‘modified’ implant, the most highly stressed area was adjacent to the point of implant insertion. On the whole, these stresses were quite well distributed over the entire implant area, and never reached critical levels [16].

2.2.2 Case II

Over the years, it has been necessary to develop appropriate orthodontic techniques in order to restore fully or partially edentulous maxillae with thin cortical bone, as has been demonstrated the quality and amount of bone plays a fundamental part in reducing strains in the bone adjacent to the implant under load [17]. The most commonly used techniques include maxillary sinus cerclage and lateral insertion [18], while sinus lifting, i.e., elevating the floor of the maxillary sinus, has recently attracted considerable attention [19]. However, this type of surgery is fairly invasive, involves a long waiting time before the implant structures can be used and, above all, leads to severe sequelae should the implant fail. Adding to all of these drawbacks is the fact that the endosteal membrane is extremely fragile, and can sometimes tear during the operation solely as a result of the patient’s breathing, even without being stressed by poor surgical procedures. As another immediate or long-term complication, the implant may shift into the sinus if there is little or no periimplant ossification.

For all of these reasons, attempts have been made to develop alternative techniques that conserve the physiological anatomical structure. The first to develop
such an alternative approach was Apolloni [20], who used a technique based on
cortical support outside the sinus area. However, the Apolloni implant has sev-
eral disadvantages, including a lack of primary stability, anchorage similar to that
provided by subperiosteal implants and the fact that the buccal mucoperiosteal flap
may be positioned in an area where it is often subject to tensile stresses from muscle
insertions.

The ‘two-phase’ implant technique is a further evolution of this method which
provides good primary stability, as the implant can be retained immediately to the
bone by means of the teeth at the outside edges (fig. 5) [11]. In addition, it is
covered by a bone graft that not only makes it a bicortical endosseous implant, but
also provides a sufficient support surface for the mucoperiosteal flap.

As this technique would appear to be less invasive and easier to implement, we
proceeded to verify whether it was also biomechanically valid. The sinus lifting
technique was compared with the ‘two-phase’ implant approach. Specifically, the
first technique considered employs conventional implants following an autologous
bone graft from the palate, while the second involves inserting screw-type rootform
implants and innovative bracket-type implants rigidly connected by a wire by means
of a synchrocrystallization process. Implants anchored with bicortical fixation were
chosen in order to increase primary stability, thus reducing micromovements and
optimizing stress distribution.

Comparing different surgical treatments is by no means easy, as there is no
universally accepted method for defining the patient’s degree of bone atrophy, and
each patient constitutes a unique biomechanical system. We thus chose to conduct
a finite element analysis, as this makes it possible to simulate different surgical
treatments on the same patient.

The geometrical models are shown in figs 6 and 7. Full osseointegration was
simulated, since all implants are surrounded by 1 mm thick cortical bone in both
models. The volumes indicated in fig. 6 were divided into 49,000 ten-node tetra-
hedron elements, while those in fig. 7 were divided into 15,000 ten-node tetrahedron
elements (simpler geometry made it possible to use larger elements). A finer mesh
was used where results indicated high stress gradients. Each model is constituted
by four different materials, whose properties are shown in table 2 [21].
To simplify modeling, all materials were assumed to be homogeneous and isotropic, even though bone is known to show marked anisotropy. A recent study demonstrates that anisotropy, when considered, can lead to a 20% to 30% rise in mandible stress levels, suggesting that careful consideration should be given to its use in finite element studies of dental implants [22]. In the authors’ opinion, considering that bone patient mechanical characteristics are not known, it is not worthy to further complicate the numerical model and this assumption should not invalidate the comparison between the two surgical approaches.
The maxillar portion was fully constrained at both ends.

Two heavily loaded situations were simulated where load is concentrated at the middle of the bone segment. In the first case, a pure vertical load of 200 N was applied, while a 40 N posteriorly-directed horizontal component was added in the second case.

Three different areas were analyzed: the external surface of the cortical bone, the transition between cortical and trabecular and implant longitudinal and cross sections.

Close attention was devoted to bone in all cases. The von Mises equivalent stress is shown in order to provide a clearer overview and to identify the most highly stressed areas.

Load is quite well distributed in both cases of load, as stresses are spread over the entire bone even if the load is applied on one single node. Dangerous stresses are reached in very small areas, viz., the bone-implant interface in the case of conventional implants (fig. 8) and the points where the retaining pins enter the bone for more innovative subperiosteal implants (fig. 9).

Though stress levels are quite similar for both surgical approaches, there are certain differences. First, the external cortical surface is subject to higher stresses when the subperiosteal implant is used (figs 10 and 11), as this type of implant sits atop the cortical bone; though no dangerous stresses are reached, this is because the geometrical model simulates a perfect fit between the implant and cortical bone.

Second, trabecular bone is not highly stressed, because most of the load is carried by the stiffer cortical bone. Some stress concentration occurs where angled...
screws exit trabecular bone and sit against cortical bone, though stress still remains low. Third, cross-sectional views show that stress levels are similar at both ends, while stress concentrations occur in the middle sections in the case of conventional implants (fig. 8).

Figure 9: von Mises stress pattern for the second case of load: subperiosteal implant design. Stress values are in MPa.

Figure 10: von Mises stress pattern in cortical bone for the second case of load: conventional implant design. Stress values are in MPa.
These stress concentrations are found at the implant’s point of emergence, at the cortical bone-grafted bone interface (fig. 9), and at the distal ends of the screws (fig. 12). This is in good agreement with Meyer et al. [17], who identified the same critical areas in the case of a solitary implant and found similar stress levels in the case of an atrophic maxilla with good trabecular bone quality. According to the same
author, larger stress should be expected in the case of poor trabecular bone quality because most load is carried by cortical bone, whereas an average maxilla has a thicker cortical layer which would result in lower stress.

3 Pelvic floor biomechanics

3.1 Question: how important is it to restore the supporting structures for pelvic organs, i.e., the urethro-pelvic and pubo-vesical ligaments, after resection surgery?

Image diagnosis of bladder conditions has long been entrusted to urethro-cystography.

This examination shows the degree of rotation and deflection of the cervico-urethral tract, and the degree of opening and closing of the neck of the bladder at rest, during micturition and under stress. In some examination centers, it has been replaced by ultrasonography to evaluate relationships between the cervico-urethral angle and neighboring structures.

Urodynamic investigation, by itself, provides analytical manometric information concerning the detrusor and its relationships with the sphincter mechanisms. It reveals bladder and sphincter dysfunction, but cannot supply a picture of pelvic support. CT and MRI are regarded as second-level examinations. They provide a 3D reconstruction of the muscles, fasciae and ligaments, both at rest and during a Valsalva’s manoeuvre [23].

At present, however, no diagnostic examination is capable of providing a true 3D dynamic picture in orthostatism and under load, together with a description of the extent of stresses and deflections of the pelvic floor [24–26].

3.2 Structural analysis and findings

A CT image from a 24-year-old woman was used to create a 3D geometrical model of the pelvic bone, muscles and organs. Mimics graphics software was employed to reconstruct the surfaces, while Rhinoceros software was used to refine the images and save them in the IGES format compatible with the pre-post processor and the finite element solver used. Geometries of different organs were saved in individual subsystems. Each subsystem (pelvis, obturator muscles, levator ani muscle, colon, uterus, vagina, bladder, urethra, pelvic fascia, sphincter, perineum) was meshed with triangular shell elements to provide a good simulation of surfaces with abrupt changes in section. The model consisted of a total of 10,084 nodes and 21,250 elements (fig. 13).

Obturator muscles with their origins on foramina and their postero-lateral insertions on femurs were first inserted in the pelvic bone system. The elevator ani group (LA) was inserted anteriorly on the pubis, laterally on the tendon of each internal obturator muscle and posteriorly on the sacro-coccygeal tract. The pelvic fascia was inserted in the central space above the LA and attached anteriorly to the pubis,
Figure 13: Numerical model of the pelvic bone, muscles and organs.

laterally to the tendinous arch and posteriorly on the sacro-coccygeal insertion. Anterior perineum, urethral sphincter, tendon center and posterior perineum with anal sphincter and ano-coccygeal ligament were positioned in an underlying plane.

These muscles and ligaments were joined to each other and to the pelvis by their own ligaments and fibrous insertions. The anterior perineum was attached to ischio-pubic rami by means of hinges, allowing it to deflect when loaded. Urethrophevic ligaments running transversely into the endopelvic fascia and pubo-urethral ligaments running longitudinally into periurethral fascia were represented.

Urethro-pelvic and pubo-vesical ligaments were reconstructed in the same way. Periurethral and perivesical parts of endopelvic fascia were inserted into the anterior segment of pubis and laterally to obturator tendons.

The vescico-urethral tract was interposed in the anterior hiatus of the pelvic fascia and the underlying perineum by describing connections between pubourethral ligaments, the pelvic fascia and the pubis. In the same way, the vagina and uterus were positioned in the openings on the pelvic fascia and perineum. Utero-sacral, cardinal pubo-rectal and utero-sacral ligaments were completed.

Since the space between the viscera is small, fasciae and ligaments were realistically represented as fused together. Lastly, the ano-rectal tract was inserted in the anal sphincter.

The sphincter was adjacent to the medial margin of the LA and connected to coccyx by the corpus ano-coccygeum.

The link between pelvic fascia and perineum was established by inserting several vertical tie-rods or struts designed to shorten when loaded and act as shock absorbers.

Each structure was assigned a specific Young’s modulus ($E$) and Poisson’s ratio ($\nu$) (table 3). These mechanical properties were obtained from experimental testing for the bladder [27], and were taken from the literature for the other structures [28–30].

All materials involved were considered to be isotropic with linear behavior.
Table 3: Mechanical properties of materials.

<table>
<thead>
<tr>
<th>Materials</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bone</td>
<td>20,000</td>
<td>0.3</td>
</tr>
<tr>
<td>Ligaments</td>
<td>1000</td>
<td>0.3</td>
</tr>
<tr>
<td>Muscles</td>
<td>0.400</td>
<td>0.4</td>
</tr>
<tr>
<td>Bladder</td>
<td>0.028–0.150</td>
<td>0.4</td>
</tr>
<tr>
<td>Pelvic fascia</td>
<td>0.030</td>
<td>0.4</td>
</tr>
<tr>
<td>Colon</td>
<td>0.025</td>
<td>0.4</td>
</tr>
<tr>
<td>Uterus and vagina</td>
<td>0.020</td>
<td>0.4</td>
</tr>
</tbody>
</table>

The subject was assumed to be in orthostatic position. Pressure and gravity loads were applied.

The average pressures exerted on bladder and colon by urine and faeces were 10 and 50 cm H₂O, respectively. A gravity load of 60 N was selected to represent the weight of the intestines and pelvic organs estimated from donor bladder measurements.

Structural analysis indicated that the highest stresses for this load situation are in the anterior tract of the pelvic fascia and the anterior perineum adjacent to the sphincter area. Some changes occurred when the urethro-pelvic and the pubovesical ligaments were not included in the model. In this simulation, the prepubic tract of the pelvic fascia deflected significantly. The load is concentrated on the underlying perineum near the urethra and sphincteric tracts that become unsteady under to excessive stress.

These findings support the validity of surgical procedures designed to reconstruct stress-relieving lines of force and bearing constraints.

4 Conclusions

Combining the analytical techniques of engineering with clinical experience, and orthopedic experience in particular, has enabled us to avoid major biomechanical errors and significantly reduce the number of short-term prosthesis failures – provided, of course, that no problems arise as a result of surgical quality or biological causes.

Our analysis of the experimental and theoretical methods used to solve biomechanical problems of a structural nature in concrete applications has led us to conclude that the only practicable approach consists of combining the finite element method with clinical verification.

We have outlined several FEM applications drawn from our experience to show how this method can make a far from negligible contribution to clinical practice.

Currently, the frontiers of structural biomechanics lie both in customized studies of biomechanical systems which take the unique biological properties and
configurations found in each individual into account, and in studies which do not stop short with the immediate post-operative period, but also investigate how the biomechanical system created by implanting a prosthesis changes over time. When the biological structure affected by surgery is bone, this type of long-term analysis necessarily involves observing and evaluating the biostructural role played by bone remodeling. Bone remodeling is the phenomenon whereby bone tissue adapts its structure, morphology and mechanical properties to changing mechanical stresses. The remodeling equations are constitutive in nature, and determine the rate of change in bone density and form in response to the mechanical stimulus that guides the structural alteration. The nature of the remodeling stimulus is not fully understood, and there are various theories linking remodeling to changes in mechanical parameters.

The objective is to investigate the changes in the structure and mechanical properties of the new bone tissue that occur as it is remodeled around the prosthesis, and determine whether the new biomechanical system thus created is capable of reaching equilibrium.

In the future, it is likely that computer aided surgery (CAS) methods will be developed which, together with increasingly sophisticated imaging techniques, will make it possible to perform customized, interdisciplinary studies of individual cases that combine preoperative planning with structural analysis and virtual surgery. There are also good prospects for using robots in surgery.

References


