Numerical and experimental evaluation of stresses in a human mandible

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ABSTRACT
In the last two decades, medical and engineering specialists developed a more and more close cooperation. Dental and orthodontic sciences represent an excellent example of this synergic view. In particular, detailed information on stress distribution in the implant region may greatly help clinical experts to reduce risk of failures thus preventing the need for re-implantations. Of course, modeling and analysis of biological/medical structures is often based on subjective information that may considerably change from patient to patient.

In view of this, the present research aims to set up a framework for building reliable analysis models from geometrical information gathered with medical imaging tools. In order to check the feasibility of the procedure, we considered the case of a human mandible for which a virtual model is reconstructed from Computerized Tomography. This case is very indicative because of the very high complexity of the anatomical district analyzed. The mandible model thus reconstructed has been studied with FEM analyses simulating four different scenarios that may occur during mastication.

The experimental part of this research is the processing of CT scan files which actually requires a very accurate study and special cares. The experimental work then prosecuted with the construction of the a stereo lithographic model that can be utilized for photoelastic investigations.

INTRODUCTION
Finite element method is an efficient tool for predicting structural behavior of complicated anatomical districts under a variety of loading conditions. CAD models given in input to FEM models can be generated directly from medical image scans acquired by powerful techniques such as Computer Tomography (CT) or Nuclear Magnetic Resonance (NMR). However, the quality of numerical results may strongly be affected by the degree of accuracy achieved in the 3D geometry reconstruction process of the anatomical districts investigated. Another important aspect to be considered is the effective capability of reproducing loading and boundary conditions. Ideally, numerical models should be refined based on experimental data gathered from in vivo tests. However, this task cannot be accomplished especially in cases where positioning measurement devices is very difficult.

More realistically, a good compromise is to have an intermediate stage where a prototype of the anatomical district is constructed and experiments are carried out on the prototype. Rapid Prototyping (RP) is a technique able to build photoelastic (PE) models from medical imaging techniques such as CT and NMR. Stereolithographic (SL) machines can read these data and build resin models even of very complicated geometry. Remarkably this capability is not limited to hard tissues. In fact, geometry of soft tissues can be reproduced at a great deal of precision as well as the position of nerves, blood vessels. Virtual reconstructions of anatomical districts may also help analysts to optimize clinical protocols or predict effects of clinical interventions in order to prevent malfunctioning thus increasing bio-compatibility. For instance, surgeons can greatly improve their understanding of CT and NMR data even though they are not familiar with this kind of analysis to the same degree as radiologists are. Finally, because SL is able to build very precise models (up to few tens of microns), RP models can be used as masters for realizing ad hoc prostheses for each patient thus preventing in vivo modifications of the implant. More recently, biocompatible acrylic resins are being studied in order to build parts directly implantable in the human body.

Photoelastic models are usually built of materials (for instance, resins) that exhibit bi-refracting behavior under mechanical loading [1] This results in the formation of a fringe pattern from which it is possible to find principal stress directions as well as the loci where the difference between principal stresses is constant. These information can be utilized in order to find zones of stress concentration. Although photoelastic models could not reproduce effective the whole set of conditions existing at the implant/bone interface (anisotropy, bone remodelling, etc.), we have to remind that stresses generated by the insertion process itself must not be too high in order to facilitate the physiological remodelling process without risks of tissue necrosis and the subsequent failure of the implant. Therefore, photoelastic measurements should be considered an effective tool for estimating bio-compatibility of implants.

In view of the previous discussion, this paper aims to develop a framework for building reliable virtual models from geometrical information gathered with medical imaging tools. In order to check the feasibility of the procedure, we considered the case of a human mandible for which a virtual model is reconstructed from Computerized Tomography. This case is very indicative because of the very high complexity of the anatomical district analyzed. The mandible model thus reconstructed has been studied with FEM analyses in order to simulate four different scenarios that may occur during mastication.
CONSTRUCTION OF THE 3D MODEL OF THE MANDIBLE

A human mandible has been subjected to CT scan. The CT data file included 72 slides each of which is 0.6 mm thick. Images have been processed using the MIMICS 7.2® program by Materialise Inc. The region of interest including the mandible district has been separated from the image background by properly setting a threshold value. Pixels for which the grey level value is higher than threshold value will be included in the spatial domain limited by the object. As is clear, the final geometry of the mandible that results from scans will depend on the threshold value. In general, threshold values may change for the different regions of the mandible. Here, we chose values according to literature. For instance, Figure 1 shows that the optimal threshold values are 1500 and 1400 respectively for the mandibular bone and teeth. The paths indicated by the red arrows clearly show local maxima in correspondence of the mandibular arch (Figure 1a) and of teeth (Figure 1b). The region growing command served to separate mandibular bone from the last two cervical vertebrae.

![Fig. 1. Distribution of grey levels: a) mandible body; b) tooth location.](image)

Although MIMICS® is able to generate and visualize a 3D domain (see Figure 2) by interpolating the 2D images obtained for each slide, the model itself cannot given as input to the ABAQUS® finite element code used in this research. Therefore, the mandible model has been encoded under a different format in order to import the model in the FEM environment.

![Fig. 2. 3D model of the mandible generated with MIMICS®.](image)

For each slice, the poly-line command has been used for tracing contours of the different bone segments acquired with the CT scan. However, poly-lines define very irregular and wavy curves (see Figure 3) that hence cannot be utilized. Therefore, poly-lines must be transformed into continuous curves defined by analytical expressions able to describe the object geometry. These curves – usually referred to as Non-Uniform Rational B-Splines (NURBS) – are defined by polynomial expressions that well approximate irregular surface contours. For each NURBS curve, the polynomial order and the number of control points (left to the user to specify) are taken as characteristic parameters. Although a NURBS curve will approximate better its corresponding poly-line as the number of control point is higher, too many control points could result in wavy lines that do not reproduce correctly the contour shape. Here, we chose the NURBS polynomial order and
control point number, respectively, as the MIMICS® default order and as the maximum control point number allowed by the software. The latter setting is motivated by the fact that the precision reached by MIMICS® becomes highest in correspondence of the maximum control point number allowed by the program.

![Fig. 3](image)

**Fig. 3.** Effect of the poly-line command limiting contours on each CT slide

In order to have a better correspondence between NURBS and poly-lines, the surface of each tooth has been reconstructed independently by creating level curves for each slide. Figure 4 show the mandibular arch before (Figure 4a) and after (Figure 4b) separating teeth from the rest of the region. Separation has been obtained using the *erase* command which deleted pixels at interfaces; later, poly-lines have been recreated.

![Fig. 4](image)

**Fig. 4.** Separation of teeth from mandibular arch.

The mandible geometry has been described by 159 NURBS curves. Fourth order NURBS have been utilized. In particular, the central zone of the mandible required 45 curves each of which included 77 control points. Each mandible ramus required 29 curves still including 77 control points each. Finally, each tooth required 4 NURBS curves with 40 control points. NURBS have been grouped into three different sets and encoded in the IGES format. Figure 5 shows the curve sets imported in the the PTC Pro-Engineer Wildfire 2.0® CAD environment. Finally, the 3D domain limited by the NURBS envelope surface has been defined by means of the *blend* feature. Since blending is successfully executed if NURBS curves have the same number of control points, we had to import three different curves at the interfaces between mandible central zone and the two rami. The same has been done in order to attach teeth to the mandible. It should be noted that only one NURBS curve every four curves has been used for reconstructing teeth because dentine resulted much more sensitive to threshold changes than the bony region into which teeth are inserted. This fact resulted in NURBS curves much more irregular than for the mandible. In addition, we have to consider that some pixels were deleted in order to facilitate separation of teeth from the rest of the model. Therefore, the degree of accuracy of
teeth reconstruction is certainly much lower than for the mandible. However, the differences in model fidelity will not affect significantly stress distributions. The model finally reconstructed is shown in Figure 6.

![Fig. 5. Pro-Engineer view of NURBS curves generated by MIMICS](image)

**Fig. 5. Pro-Engineer view of NURBS curves generated by MIMICS**

![Fig. 6. Frontal view (a) and lateral view of the reconstructed 3D model of the mandible](image)

**Fig. 6. Frontal view (a) and lateral view of the reconstructed 3D model of the mandible**

**FINITE ELEMENT ANALYSIS**

**Model**

The solid model of Figure 6 has been imported into the ABAQUS® software [2]. In order to load the CAD model, the mandible has been divided in 17 parts: the central region, the two rami and the 15 teeth. The TIE command allowed us to merge the different segments. The 3D model has been meshed with 4-node C3D4 tetrahedral elements that are well suited for meshing irregular and complex geometries. Convergence analysis has been carried out in order to refine mesh opportunely. The FEM model shown in Figure 7 finally included about 80000 elements and about 99000 nodes.

![Fig. 7. Finite element mesh of the mandible](image)

**Fig. 7. Finite element mesh of the mandible**
Materials
The FEM model of the mandible district shown in Figure 7 included two different materials: the mandibular bone and the teeth. Materials have been modelled as homogeneous and linearly elastic. The Young's modulus and Poisson's coefficient were, respectively, set as 13.5 GPa and 0.3 for the mandibular bone. Teeth have been modelled as entirely made of dentine whose elastic modulus is 18.5 GPa and the Poisson’s coefficient is 0.4. Effect of dental pulp was not considered. [3]

Boundary conditions and loads
The presence of condiles has been simulated by defining two reference points (RP) at the locations of the temporomandibular joint. The reference points so defined have hence been connected to the mandible arms using the coupling constraints available in ABAQUS. Because the temporomandibular joint disc Young’s modulus is typically about 6 MPa (Beek et al., [4]), reference points have been constrained to three fixed points by means of springs directed as the coordinate system. Coupling constraints enforce model surfaces to have the same displacements of the reference points to which they are connected. Therefore, mandible arms can rotate about the line passing through the reference points and even translate in the directions 1,2,3.

The mandible, hinged at the condiles, is subjected to mastication that has been simulated including in the FEM model the presence of the following muscles: superficial masseter (SM), internal masseter (IM), internal pterygoid (IP) and external pterygoid (EP), anterior-temporal (AT) and posterior-temporal (PT). Each muscle has been modelled by coupling a reference point outside the 3D model and the surface portion where the muscle itself is inserted into the mandibular bone. The resultant force exercised by each muscle has been applied at the corresponding reference point. Force amplitude and direction have been set equal to those measured by Faulkner et al. [5] that estimated the effect of muscular forces by combining electromiographic data and measures of muscle sections. Table 1 reports for each muscle included in the FEM analysis the resultant force amplitude, the force components along directions 1-3 and the corresponding direction cosines.

<table>
<thead>
<tr>
<th>Muscles</th>
<th>Direction cosines of resultant muscular forces</th>
<th>Magnitude of forces developed by muscles [N]</th>
<th>Components of muscular forces [N]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Direction 1</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Right arm</td>
<td>-0.1</td>
<td>0.76</td>
<td>0.64</td>
</tr>
<tr>
<td>PT</td>
<td>-0.07</td>
<td>-0.34</td>
<td>0.94</td>
</tr>
<tr>
<td>AT</td>
<td>0.32</td>
<td>-0.03</td>
<td>0.94</td>
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<tr>
<td>IP</td>
<td>-0.25</td>
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<td>-0.25</td>
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<tr>
<td>EP</td>
<td>0.27</td>
<td>-0.15</td>
<td>0.95</td>
</tr>
<tr>
<td>SM</td>
<td>0.27</td>
<td>0.18</td>
<td>0.94</td>
</tr>
<tr>
<td>IM</td>
<td>0.1</td>
<td>0.76</td>
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<tr>
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</table>

Four different working conditions were considered where occlusion takes place at different positions along one arm of the mandible: 1st premolar, 2nd premolar, 1st molar and 2nd molar (see Figure 8). In order to realize this scenario, the vertical displacement of the upper surface of teeth at which occlusion occurs has been constrained to be zero.

Table 1. Amplitude, direction and components of muscular forces on mandible

Fig. 8. Occlusal loads acting on the mandible
NUMERICAL RESULTS

The linear analysis carried out for the mandible using ABAQUS Standard® revealed that stress concentrations occur in the lower side of the mandible. For instance, Figure 9 shows two views (from the labial and the buccal side) of Von Mises stress distribution when the occlusion takes place at the 2nd molar. Since the occlusion is realized only in one arm of the mandible, the resulting stress field is not symmetric. The mandible exhibits stress concentrations in the segment between each condile and dental arch. Stress peaks are however below 8.5 MPa.

Figure 10 shows that the position of the occlusal surface does not affect significantly stress distribution in the mandible arm opposite to the occlusal region. Stresses are higher in the unoccluded mandible ramus. Moreover, it can be seen from the figure that Von Mises stress peaks in the occluded ramus quickly decrease as the occlusal surface moves away from condilar region and towards the anterior region of the mandible.

Figure 11 shows force reactions developed at the interfaces where the occlusion takes place. Again, force reactions decrease as the occlusal surface moves towards the anterior part of the mandible. This behaviour can be explained with the argument that as the distance from condyles increases the lever arm of the occlusal force also increases while muscular actions stay always the same. In order to have equilibrium, the reaction force will decrease as its lever arm increases, that is as the occlusal surface moves away from condiles.
DISCUSSION
This paper presented an example of reconstruction and analysis of a complicated biomechanical district. A 3D model of a human mandible was built starting from CT scan data and then analyzed using a general purpose finite element code. In order to complete the process, experimental tests should be carried out. Since it is very difficult to measure mandibular stresses, in vivo prototypes of the mandible made of photoelastic resin can be realized by Stereolithography. To this purpose, Figures 12a and 12b respectively summarize the SL process and show the VIPER SI2 SL device used in this work.
As is known, Stereolithography is a technological process which serves to build 3D resin made models [6]. A dedicated software divides a CAD model into slices thus generating a series of very thin layers. The resulting cross sections are coded in fashion of files readable by the SL system. The typical SL system arrangement in Fig. 12a includes [7]:

- a workstation, through which it is possible to control the entire model construction process;
- a solid state UV laser which can move in the XY plane;
- a set of optics for steering, orienting and focusing properly the laser beam;
- dynamic mirrors for scanning along X and Y axis;
- a vat containing liquid photopolymer (epoxy resin);
- a platform;
- an elevator which makes platform move along the Z axis;
- a computer controlled recoating system.

The model building process includes six basic stages:

1. The laser beam is focused onto the vat surface.
2. The laser spot moves through the vat cross section following trajectories coded in the primary CAD file. As the laser beam hits vat surface, a chemical reaction occurs causing polymerization of the liquid monomer and hence the creation of a solid layer.
3. The elevator moves down the part constructed in stage 2 by approximately one layer thickness below the free resin surface.
4. The recoater blade then sweeps again resin surface before the next layer is drawn. This is done in purpose to apply a fresh coating of liquid resin to the part.
5. Stages 2-4 are repeated until the whole prototype is realized.
6. The finished product is put in a UV furnace for a post-curing treatment.

Figure 13 shows the stereolitographic model of the mandible realized by giving in input to the SL machine the CAD file of the FEM model. The model is made of RP Cure 400 ND epoxy-resin. The high fidelity reached in reproducing the CAD model suggested to carry out photoelastic analysis in order to individuate critical regions possessing stress concentrations. Consistency between FEM predictions and photoelastic data is ensured by the fact that we carried out a linear FEM analysis. Therefore, numerical and experimental results exhibited the same trends while stress values are proportional to values of material properties adopted for the photoelastic resin and the mandible regions.

In view of this, we can conclude that SL is an effective tool for speeding up analysis of complicated biomechanical districts. Medical imaging, CAD, SL and photoelasticity should therefore be integrated in order to have an automatic process that allows us to by-pass in vivo tests or, at least, reduce significantly their duration. Although it might be argued that anatomical districts may change from patient to patient, the degree of fidelity achieved by rapid prototyping models however allows us to deal with results distributed on a statistical basis.

REFERENCES