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The Role of Finite Element Analysis in Studying Potential Failure of Mandibular Reconstruction Methods

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Additional information is available at the end of the chapter

<http://dx.doi.org/10.5772/64890>

Abstract

Defects of the mandible occur after trauma or resection after infection or tumours. There have been many methods espoused, but many methods can fail especially if the biomechanics of the mandible is not considered fully. As the only moveable, load-bearing bone of the skull, the mandible is subject to loads and stresses unique to it due to its shape, location and function. This chapter reviews the basic knowledge of the mandible necessary to perform finite element analysis, the challenges and then reviews several studies that have been done. The authors' personal research is detailed to illustrate how finite element analysis can be used to look at potential failure of a new method for mandibular reconstruction and implant evaluation.

Keywords: mandible reconstruction, finite element analysis, potential failure

1. Introduction

The lower jaw or mandible is the only load bearing, moveable bone of the skull. Defects in the lower jaw or mandible can happen as a result of trauma, infection or after a resection for pathology, which can be benign or malignant. The field of medicine or surgery has never been able to satisfactorily reconstruct a mandible after the original tissue has been gone. An unrepaired defect in the mandible, depending on the location of the defect would lead to (1) constriction of the remaining tissue around the defect leading to a malocclusion due to the pull of the muscles, (2) collapse of the arches leading to inability to eat and function and (3) difficulty swallowing or even breathing especially when lying flat due to loss of attachment of the tongue, sometimes leading to aspiration or asphyxiation.

2. Methods for mandibular reconstruction

There have been many methods advocated for reconstruction of the mandible, most of which have never fully considered the biomechanical forces acting on the mandible both in the short and long term. These methods have been “tried”, most times with catastrophic results. Among the methods for mandibular reconstruction advocated are:

- 1) Soft tissue flap
- 2) Autogenous bone blocks
- 3) Mandibular reconstruction/bridging plate
- 4) Cancellous bone in titanium mesh
- 5) Vascularized free flap
- 6) Newer methods like the endoprosthesis or alloplastic replacement
- 7) Tissue engineered bone scaffold

The current gold standard is still the vascularized free flap, which needs a long operation time and harvest of tissue from another part of the body [1, 2].

This brings the question of “what is the ideal method of mandibular reconstruction?” to the fore.

2.1. Ideal method

The ideal method of mandibular reconstruction would:

- 1) Reconstruct the missing soft tissue and bone
- 2) Allow replacement of teeth
- 3) Not need a long surgery
- 4) Easy to learn without needing extensive training and skills that are hard to learn
- 5) Not need to take tissue from another part of the body
- 6) Cost effective
- 7) Not need prolonged hospitalization and recovery
- 8) Allow the patient to eat and function early or immediately
- 9) Be able to withstand the forces of biting and chewing permanently or for a long time

No such ideal method exists.

2.2. Challenges to mandibular reconstruction

The challenges to replacing a section of the mandible lie in its form and function as well as its unique location in the body. Any hardware (plates and screws) used to fix the mandible would

undergo unique stresses not seen in other parts of the body. There is a non-axial load in that the long axis of the teeth is about perpendicular to the long axis of the mandible. The overlying tissue in the mouth is thin; any break in the soft tissue could lead to exposure of the hardware to the bacteria and saliva in the mouth, leading to the formation of biofilms. Ingestion of hot and cold food and liquids can cause expansion and contraction of the hardware, which is dissimilar to that of the underlying bone.

3. Mandibular biomechanics

The biomechanics of the mandible has not been very well studied. Mandibular biomechanics can be studied with finite element analysis, strain gauges or photoelastic models. Finite element analysis allows study of the forces throughout an entire structure but is limited to static forces. An understanding of the anatomy and biomechanics of the intact mandible is necessary prior to looking at how to conduct a finite element analysis of the mandible.

3.1. Anatomy of the mandible

The mandible is a U-shaped bone, connected at the temporomandibular joints at both ends to the skull. It consists of a corpus (body), the symphysis, which connects both right and left at the midline, the alveolar bone, which supports the teeth, the ramus with the condylar as well as the coronoid processes (**Figure 1**).

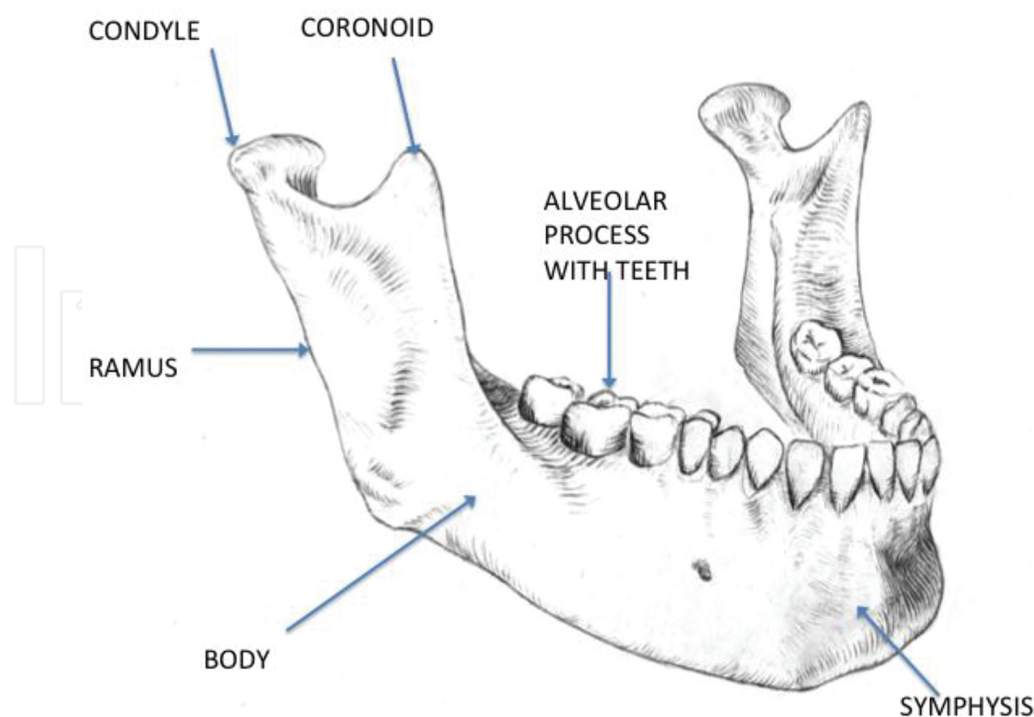


Figure 1. Schema showing the gross anatomy of the mandible.

The teeth are connected to the alveolar bone by periodontal ligaments, which act as a gomphosis (allowing minute movement). These ligaments are inserted into the bone on one end and into the cementum of the teeth.

The muscular attachments can be divided into three groups:

- 1) Muscles of mastication—medial pterygoid, lateral pterygoid, masseter and temporalis; only the lateral pterygoid muscle assist in opening the mandible, the rest closes the mandible.
- 2) Suprahyoid muscles that assist in some opening of the mouth and swallowing—mylohyoid, hyoglossus, genioglossus, digastric muscles.
- 3) Muscles of facial expression that insert into the mandible—buccinators, depressor anguli oris, mentalis.

The outer surface of the mandible consists of dense cortical bone, the thickness of which varies. In certain areas, there is only cortical bone throughout. The alveolar processes and in the middle of the mandibular body and part of the ramus consist of cancellous bone (bone marrow). There is a nerve coursing through the mandible in a canal, which usually does not play a role in terms of biomechanics.

The attachment of the mandible to the skull consists of the attachments of the muscles of mastication and the temporomandibular joint. The temporomandibular joint consists of two joint spaces, the superior and inferior joint spaces, surrounded by a capsule consisting of elastic collagen fibres and divided by the articular cartilage, which is a fibrocartilage. The movement of the joint consists of two distinct movements, which are Phase I (rotation) about a hinge for the first 20 mm followed by Phase 2 (translation), which is affected mainly by the action of the lateral pterygoid muscle pulling the entire condyle to the front and out of the glenoid fossa onto the part of the zygomatic process of the temporal bone. The forward movement of the condyle is limited somewhat by a protrusion called the articular eminence.

3.2. Biomechanics of the intact mandible

The mandible functions as a Class III lever. During function, there is a zone of tension on the alveolar part of the mandible and a zone of compression on the lower border. Studies by Meyer [3, 4] showed bone deformation in the mandibular condyle region, with tensile stress along the anterior ramus as well as along the sigmoid notch area and compressive stress along the posterior ramus border. This suggested that there is a tendency of the mandible to straighten during function. This somewhat simplistic model holds true when there is bilateral and equal function and bite forces (**Figure 2**).

Upon contraction of the muscles of mastication, the mandible is bent in a sagittal plane; this is produced by the vertical component of the muscle forces, the joint reaction forces and the reaction forces from chewing motions. During asymmetrical loading (biting on one side), the largest shear forces happen between the bite force and the muscle force on the working side (the biting side) and between the muscle force and joint force on the balancing side (non-biting side). This produces then a converse load distribution with a zone of tension on the lower

border and a zone of compression on the alveolar portion in the working side and vice versa on the balancing side. This means that during incisal biting (biting on the front teeth), there is an equal amount of sagittal bending on both sides but a different deformation on the working and balancing sides during molar biting [5–7].

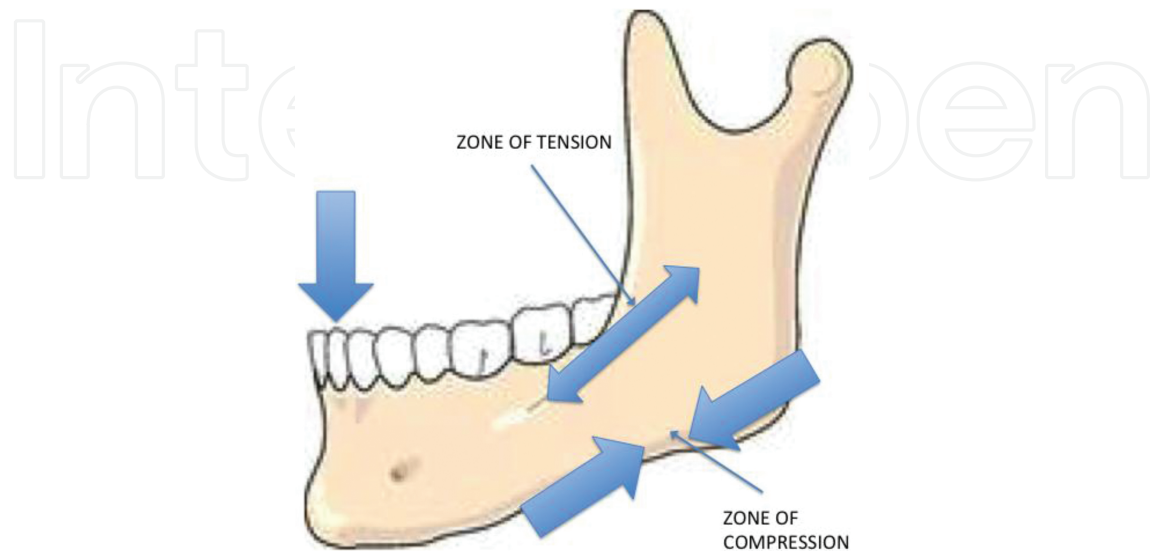


Figure 2. Zones of tension and compression in the mandible.

There is also a tendency for narrowing of the mandibular arch from parasagittal and transverse deformation upon clenching and incisal biting. This is caused by bilateral torsion of both mandibular bodies and bending at the symphyseal region leading to compression at the superior margin of the symphysis and tension at the inferior margin.

Hylander showed that the mandibular symphysis undergoes three distinct patterns of stress and deformation, that is, corporal rotation (relative outward rotation of both halves of the mandible), medial convergence (change in mandibular width during function) and dorso-ventral shear (movement of both mandibular halves relative to one another in the vertical plane) [8] (**Figure 3**).

There is also some difference in the deformation between the outer (buccal) surface of the mandible and the inner (lingual) surface. Lateral transverse bending occurs and the bending moment increases from back to front during the late power stroke of biting/clenching. The maximum magnitude of the bending occurs near the symphysis. This bending produces compressive stresses at the buccal cortex and tensile stress at the lingual cortex. The deformation has been calculated to be as large as 0.6 mm in a simulated molar bite of 526N using finite element analysis [6]. The mandible deformed in a helical pattern upwards and towards the working side, with regions of high tensile stress (15–25 MPa) from the coronoid process and ramus towards the lingual side of the symphysis. The highest value of compressive stress (15–25 MPa) was at the bite point and bilateral sigmoid notches, at the working side angle and in an area from the posterior surface of the balancing side ramus running to the lower border of the body till the symphysis. This then runs up to the buccal side from the inferior until the bite

point. Overall, the shear stresses were larger on the working side with the exception of the balancing side condyle (peak shear stress of 25 MPa). In a nutshell, this means that the mandible changes in dimensions during function as a result of its shape and muscle pull with the greatest change in dimension and deformation in the midline.

What does all this information mean and what is the practical application? It means that application of any hardware to the mandible must take into account these forces and change in dimension. This has led to the creation of Champy's ideal lines of internal fixation for fixation of a mandibular fracture [9] (**Figure 4**). Placement of bone plates and screws in the area between the zones of tension and zones of compression will tend neutralize the forces and stabilize the bony fragments enough for healing. This however uses the principle of cross bracing and load sharing, that is, using inter-fragmentary bone friction to help stabilize the bony segments.

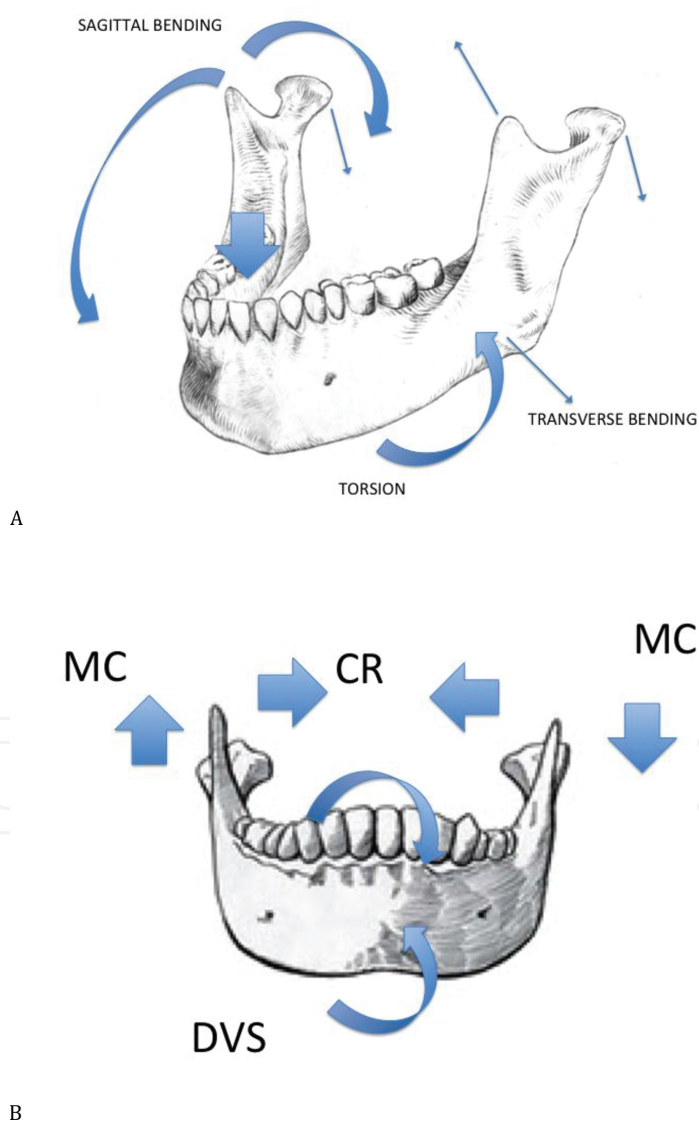


Figure 3. (A) Tendency for mandible to straighten as well as undergo torsion. (B) Forces acting about the midline of the mandible. CR – corporal rotation; MC – medial convergence; DVS – dorso-ventral shear.

In comminuted fractures or in a segmental mandibular defect, the principle of load sharing cannot be applied. We are then dependent on using load bearing bone plates, which means the material strength of the plate is the only thing keeping the mandibular segments together. The method of fixation of the bone plate to the bone also plays a factor.

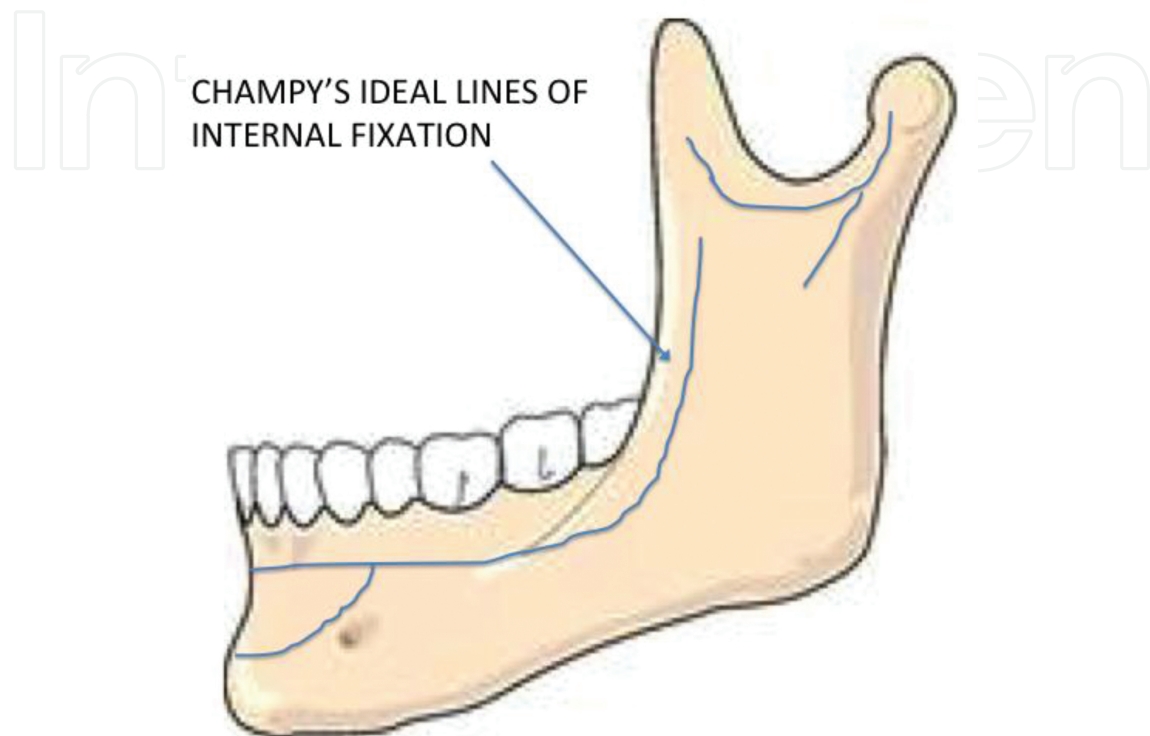


Figure 4. Champy's lines for ideal internal fixation – placement of bone plates along these lines will stabilize the bone fragments enough for healing of fractures.

4. Finite element model of the mandible

A very accurate finite element model is practically impossible to create due to the complex anatomy of the mandible. A true to accurate model that takes everything into account would need several supercomputers. Assumptions will have to be made to simplify the model and reduce the need for computing power.

Several authors have created finite element models of the mandible, each with different levels of complexity [10, 11]. Following are the steps taken to create an accurate finite element model [12].

4.1. Creation of a 3D mesh

The information about the external shaped of the mandible is needed to be able to create a mesh to input into the FEA software. Several possible methods that have been used to get the geometry of the mandible:

- 1) Digital creation of 3D model—not accurate representation of the anatomical detail. The question then arises: “At what level of intricacy would there be detriment to the level of accuracy?” Is it necessary to recreate all the intricacies (concavities, canals)?
- 2) Conversion of digitized slices or sections of a human mandible into a whole 3D structure [13].
- 3) Computer tomographic (CT) scans of a human mandible or mandibular equivalent and conversion of the radiographic images in Digitised Communication in Medicine (DICOM) format into a 3D structure in Standard Tessellation Language (STL). This information is then meshed with any number of software programs. This is the most popular method. Some authors have also used the cone beam CT, which is another way to get the CT images with a cone beam instead of a fan beam. This tends to produce many sharp triangles, which need to be simplified prior to mesh creation.

Once the 3D geometry has been obtained, it is subdivided into a finite, large number of geometrically simplified elements, connected together at the nodes. The mesh is a contiguous collection of these simple-shaped elements. Most FE software have automated mesh generation features, creating relatively dense meshes, which can be refined in different regions.

4.2. Input of material properties and boundary conditions

The material properties of the elements, namely the elastic modulus and Poisson’s ratio must be defined. Since the mandible consists of cortical and cancellous bone together with the teeth, several assumptions need to be made to simplify the model further. It is well known that bone is anisotropic in different dimensions. For purposes of simplifying the calculations, most authors have tended to assume that bone is isotropic and that the mandible is purely cortical bone. Any teeth present, which in real life would have periodontal ligaments and allows minute movements would tend to be assumed to be ankylosed to the bone, that is, fused to the bone. The teeth are composed of enamel on the outer surface of the crown followed by dentine and the pulp (a hollow cavity which contains the nerve fibres and blood supply). The outer surface of the root is covered by cementum. All these tissues have different material properties. The values for the material properties have already been determined by studies. Some errors abound when it comes to the values of the cancellous bone. Most studies have tended to remove a block of cancellous bone and then subject the block to mechanical testing to ascertain its material properties. There is evidence to suggest that this is not entirely accurate. Misch et al. (1999) showed that the presence of cortical bone increases the elastic modulus of cancellous bone [14]. When the cortical bone was present, the elastic modulus ranged from 24.9 to 240 MPa (mean 96.2 MPa). When cancellous bone only was tested, the elastic modulus reduced dramatically (3.5–125.6 MPa). This means that the values from the literature for cancellous bone are not accurate.

Boundary conditions are important to prevent movement of the individual units so that the model can be loaded and deformed as a rigid structure, allowing computations to be performed. It can be divided into essential boundary conditions (displacement constraints to anchor the model and the non-essential boundary conditions or loading conditions, which are

the forces to be applied to the model). Decisions will need to be made also about the insertion of the muscle forces and the force of each muscle. Some muscles like the masseter have three distinct types of fibres with different vectors. Most of these values are already in the literature.

4.3. Solution

With all the information, the completed model is then solved to obtain the displacements and the resulting stress and strains. In biomechanical models, what is most often sought is information on the stresses and strains, as the force is usually known.

The external forces $\{F\}$ and the mechanical properties/geometry $\{K\}$ are used to solve the nodal displacements $\{D\}$. With the nodal displacements known, the displacement fields are then interpolated from the nodal values using standard interpolating polynomial functions. The strain distribution is the differentiation of the displacement field yields and the stress distribution is then determined mathematically.

4.4. Validation and interpretation

Validation can be performed by evaluating the precision and accuracy of the model. Precision, defined as how close the model's results are to the exact solution to the biomechanical model, can be ascertained by conducting a convergence test where meshes of different refinements are created and the strains/stresses at specific locations are compared. Most reported studies have tended to use precision studies as a measure of validation as it is difficult to affix strain gauges to the human subject to test for accuracy for ethical and practical purposes.

5. Finite element analysis of the various mandibular reconstruction methods

There have been very few finite element analysis conducted on reconstructed mandibles, mainly due to this being a field that is not very well understood by the people who operate in this area. In the field of orthopaedics, the biomechanics of the limbs has been well studied and there are numerous studies using finite element analysis. This section reviews the few studies that have been conducted followed by the authors' own studies. By necessity, it is impractical to go into very much detail for each study. We will concentrate on two studies in detail at the end of this section.

5.1. Marginal resection of the mandible

Marginal resection of the mandible is performed for tumours that affect the alveolar process of the mandible but does not extend to the mandibular basal bone. One of the complications that can occur is fracture of the remaining portion of the mandible, usually from the corners of the resection where areas of stress concentration occur (**Figure 5**).

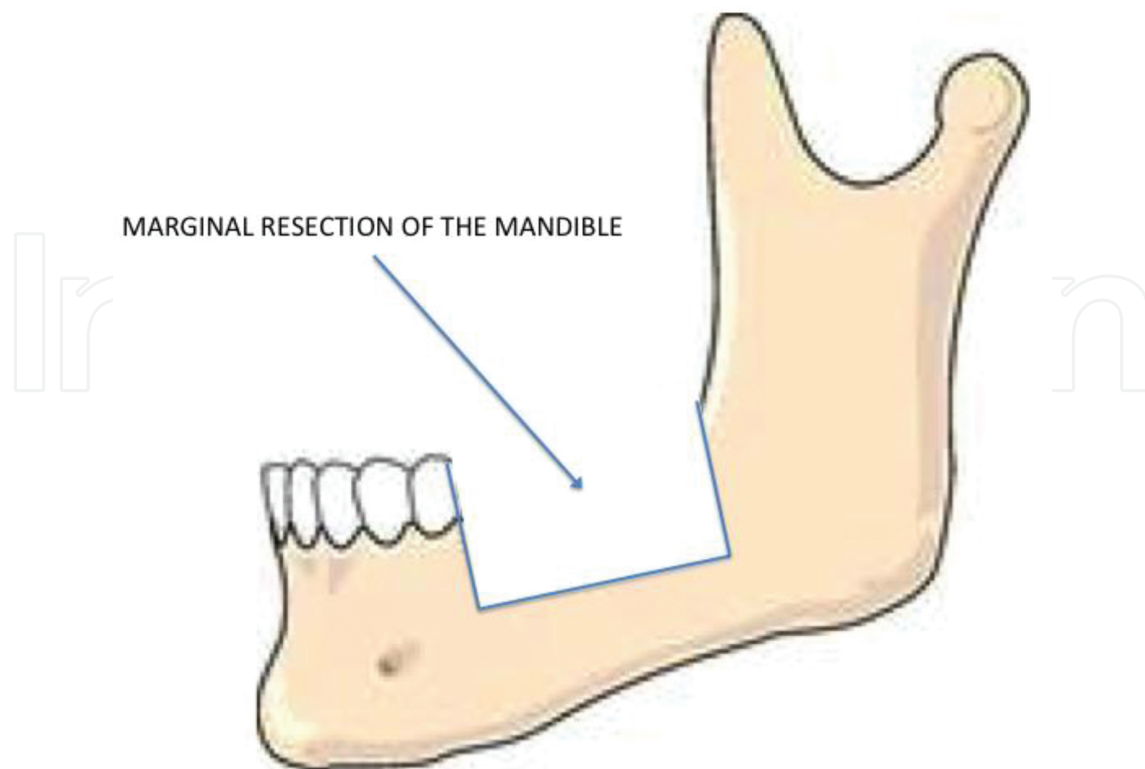


Figure 5. Marginal resection of the alveolar portion of the mandible, leaving the basal bone intact.

Wittkamp et al. [13] conducted studies on prevention of mandible fracture after a marginal resection of the mandible. The authors sectioned a human mandible, photographed the slices and digitized it. Marginal resections of different radii were placed in digitally and the areas of maximum stress concentrations in the corner of the resections were compared. It was found that an enlarge radius of resection at the corners offered the best resistance to minimize fracture after resection.

5.2. Reconstruction plates for bridging mandibular defects

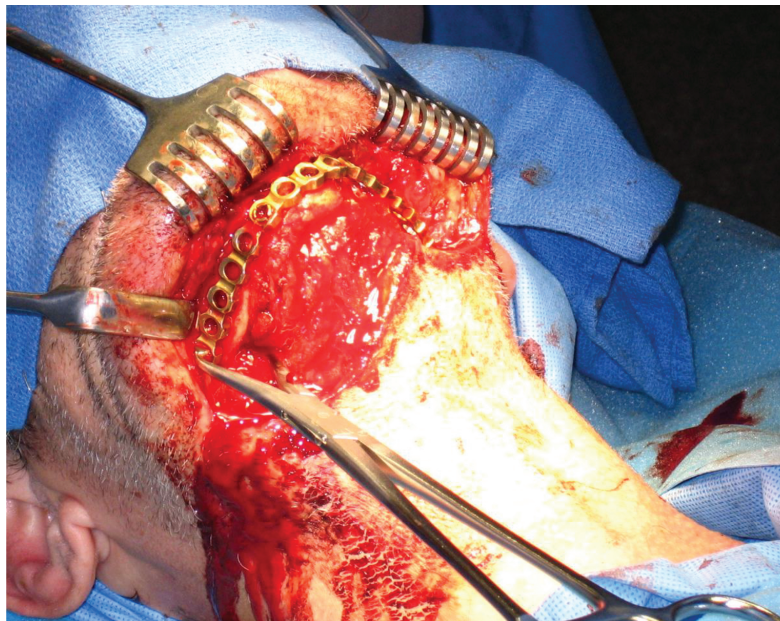
Reconstruction plates alone are sometime used to bridge a defect to maintain the space and contours of the jaw in patients in poor health or with advanced tumours. A patient might not be fit for a long surgery or it might not be worthwhile to subject a patient to a long surgery if the prognosis is poor. Sometimes also the reconstruction plate is placed together with bone grafts, which is then subject to other considerations in terms of biomechanics (**Figure 6**).

The most common complications in using a reconstruction plate to bridge a defect are plate fracture, sometimes after a few years, loosening of the screws and exposure of the plate (dehiscence) either in the mouth or through the skin.

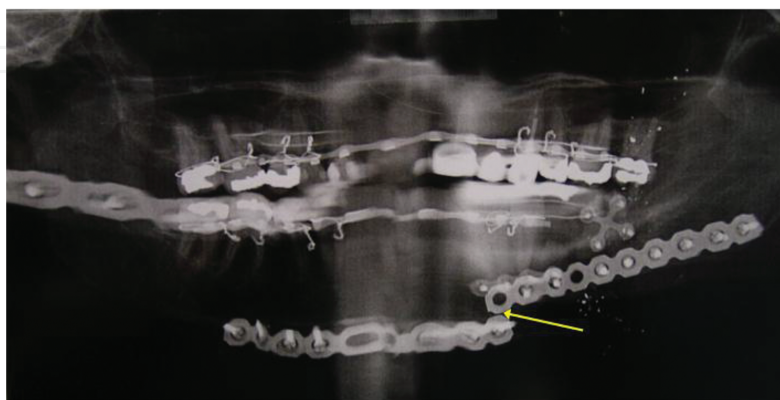
For pure mandibular bridging plates alone, the size and location of the defect and whether the defect crosses the midline (symphysis region) plays a large role in terms of complications [11, 15]. Recall the earlier section where it was shown that the midline is subject to many different

forces of tension, compression and torsion with changes in dimensions. The masticatory loads on the plates cause vertical discrepancies that can lead to bone resorption and screw loosening. Arden et al. [16] reported that defects larger than 5 cm of bone length are associated with a high complication rate as high as 81% when plates alone are used to repair lateral defects.

Martola et al. [17] hypothesized that residual stresses from bending a stiff plate (sometimes repeatedly) to adapt to the jaw contour can be a main reason for plate fracture. This makes sense from a material science point of view, in that repeated bending of a metal results in work hardening of the metal, sometime with creation of small micro-cracks which may affect the mean stress in fatigue loading.



A



B

Figure 6. (A) Reconstruction plate bridging a defect in the anterior. (B) Fracture of the reconstruction plate, a common complication. Indicated by arrow.

Kimura et al. [18] investigated the most suitable method in dispersing stresses around the screws in plate fixation to the remnant mandible after a resection of the mandible. The authors took CT scans of dry human mandibles and created eight digital edentulous (no teeth) mandible models. Defects were created on the models in the front (midline) and lateral areas and the plates were drawn onto the defects with different screw configurations. A dental implant was drawn into the opposing (contralateral) side of the defect. The material properties (based on reported data) of all the components in the model were defined: cancellous bone, cortical bone and titanium. A maximum bite force of 300 N in a vertical load pattern was chosen and used. In the analysis, the stresses were concentrated around the implant, the screw closest to the defect on both sides (crucial screw) and the plate on the non-loaded side. Defects in the midline (central defects) placed greater maximum stress on the screws. If three screws were placed for lateral defects, there was greater stress on the crucial screws, which could have been due to bowing of the plate.

A partially dentate mandible of a cadaveric male was used to create a model with defect in the front and back. The authors (Schuller-Gotzburg et al, [19]) studied the effects and change in bone stress after caudal and buccal placement of placing a bridging plate. In this study, bridging plate alone in a conventional placement, bone grafts fixed with small miniplates to the remnant mandible and also placement of a bridging plate in a caudal and then buccal location for the reconstruction. The load was 50 N on the right mandibular second premolar. The conclusion was that there would be a better biomechanical advantage and lesser stress with the plate in a caudal position.

Knoll et al. [20] investigated replacement of angle defects with a standard 2.7 mm reconstruction plate with a linear screw configuration. An edentulous finite element mandible model was created; a defect placed in the right angle with a virtual bone plate placed and the model was loaded with 135 N of force in the front. The model showed that the stresses were far in excess of the material strength of titanium and cortical bone. This result showed that there was a high possibility of plate fracture, bone loss and screw loosening. The recommendation was to redesign the plate to allow screw placement in a triangular or square configuration to further maximize the interface between bone and plate.

5.3. Vascularized free flaps

Vascularized free flaps are flaps of soft tissue and bone, harvested from another part of the body with its own arterial blood supply and venous drainage. The flaps are then placed into the part of the body that needs it, connected to the local blood supply and then fixed to the surrounding bone and soft tissue. For defects of the mandible, the two most common flaps used are the fibula and iliac crest free flaps. This is still the gold standard after a resection.

Tie et al. [21] constructed finite element models to study which flaps, the fibula or iliac crest free flaps would be best biomechanically to replace a segment of the mandible. The authors scanned the mandible, iliac crest and fibula of a healthy 30-year old volunteer and used the images to create a finite element model of the mandible. Defects were created in the anterior and lateral mandible; the outlines of the defects outlined and extracted using the software and the volume of the fibula and iliac crest made to fit the defects. The stress distribution for the

iliac crest was found to be similar to that of the intact mandible. The fibula reconstruction, however, had greater stresses (compressive and tensile) at the grafted bone with the maximum stress at the interface between native and grafted bone. The increased binding or interface between the iliac crest and native bone contributed towards better transmission of the bite forces. Their conclusion is that for smaller defects, the iliac crest would be better due to the above findings. As the fibula has greater length, it would be more suitable for larger defects.

5.4. Endoprosthesis/customized implant

An endoprosthesis is an implant, mostly titanium that is placed to replace a defect in the line of the remaining bone stumps. It is usually attached to the bone stumps with a stem, which is usually cemented or press fit. This concept has been used in orthopaedics in the long bones with great success for decades [22–25].

The concept of using an endoprosthesis for mandibular replacement was introduced by Tideman, initially as proof of concept and with several animal studies [26–28]. The decision was made to look into the mandibular endoprosthesis as a modular format rather than customized. The reasons for this were as follows:

- 1) Parts machined in large quantities are cheaper.
- 2) Customized implants require time for manufacture and are more expensive. Between the time from the scanning of the patient to the time it takes to design and manufacture as well as transport time, it may take a few weeks; this might have allowed time for the lesion if it is cancerous to grow larger in size and thus rendering the customized fit potentially inaccurate.
- 3) A stock endoprosthesis, which comes in different lengths and can be assembled in modules, allow variations and more flexibility during the surgery to adapt to defect length changes.

The modular endoprosthesis has been used, again with great success in the field of orthopaedics and musculoskeletal surgery.

The difference between the long limbs and the mandible has already been discussed previously. There is also the addition of the curvature of the mandible in the anterior region, which varies between individuals, making it difficult for a stock endoprosthesis.

Nevertheless, the animal experiments, which largely were conducted in the monkey model yielded interesting results. There were two designs: (1) the mandibular body replacement and (2) the condyle replacement. The condyle replacement had no problems with loosening and infection. The body replacement design had persistent problems with loosening between the module connections, causing infection and loosening. The cemented stems had no problems; however, a decision was made to investigate the biomechanical forces that acted on the entire reconstruction for the body replacement design [29].

The design of the endoprosthesis was changed as follows [30]:

- 1) The stem was changed from cemented to screw retained, which was to be screwed into the marrow part of the mandible.
- 2) The module connection was changed to a male and female part in a dovetail fashion, which was connected by a screw. A slight movement of 0.1 mm was designed as a stress breaker between the connection of the male and female part.

This new design was then tested for the following aims prior to any animal or human testing:

- 1) Experimental evaluation of the new design to look at fatigue performance and failure patterns by mechanical testing. The entire setup was also investigated with a finite element analysis to see if the model correctly predicted the site of failure in the mechanical testing.
- 2) Finite element analysis of the new design in a simulated human mandible under certain conditions. The defect was then to be made bigger and the stem length was to be changed in length to see what happens to the stress distribution to better predict the ability of the reconstruction to withstand failure (if the stress in any part is more than the material strength of the material) and also the location of failure.

The new endoprosthesis design was made to fit the dimensions of a human mandible and underwent mechanical testing in a jig, mounted on a synthetic mandible, which had similar elastic properties to cortical bone. The methods of mechanical testing depend on the question asked. It can range from simple three-point bending test, compressive and tensile strength to complex tests for load to failure or fatigue loading. The dimensions of the assembled endoprosthesis were 18-mm stem length with 4-mm diameter and body dimensions of 15-mm length, 16-mm high and 8.5-mm thick (**Figure 7**).

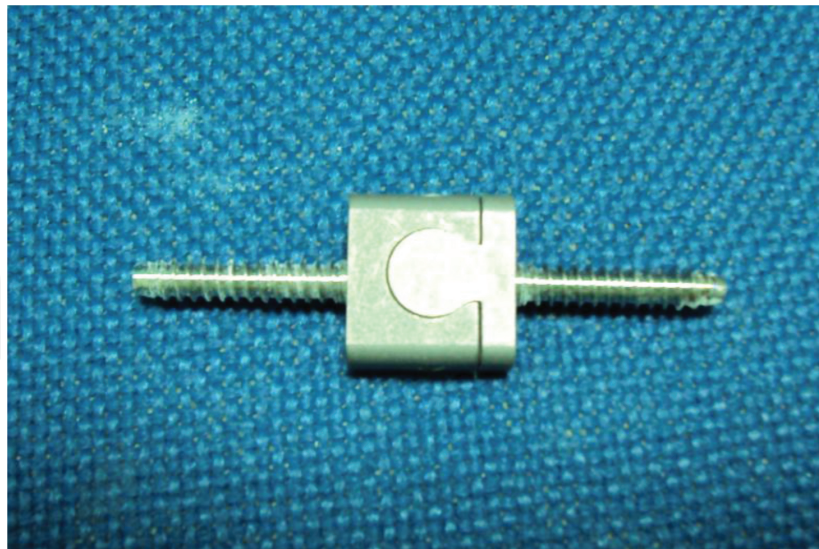


Figure 7. The endoprosthesis consisted of two screwed stems, which connected to each other in a dovetail. This was locked together by a central screw, which is inserted from the top.

Static testing revealed a tendency for the screw stem to pull out of the substrate of the synthetic mandible. Cyclic testing was then performed for up to 500,000 cycles. This revealed a tendency

for the endoprosthesis to fail with fracture or bending at the superior surface of the stem but with no loosening of the module connection, which had plagued the earlier animal experiments.

The line drawings of the endoprosthesis from the manufacturer were imported into Abaqus v6.10 (Simulia, Dassault Systemes, France). The stems were modelled as smooth cylinders to simplify calculations. Rectangular cuboids were modelled as the synthetic mandibles and bores made for stem insertion. The stems of the endoprosthesis were perfectly tied to the bores of the holes as well as the central connection screw. The cuboids were assigned the elastic properties of cortical bone and meshed with linear tetrahedral elements. It was assumed that the bone was isotropic. A bolt load of 10 N was applied to simulate tightening of the central screw. A downward force of 150 N (calculated at 80% of average static load to failure of 185 N, this was the maximum force used for the fatigue testing) was loaded on to one end of the reconstruction while the other end was given fixed boundary conditions. The load was kept constant to identify peak stresses, which lead to fatigue failure (**Figure 8**).

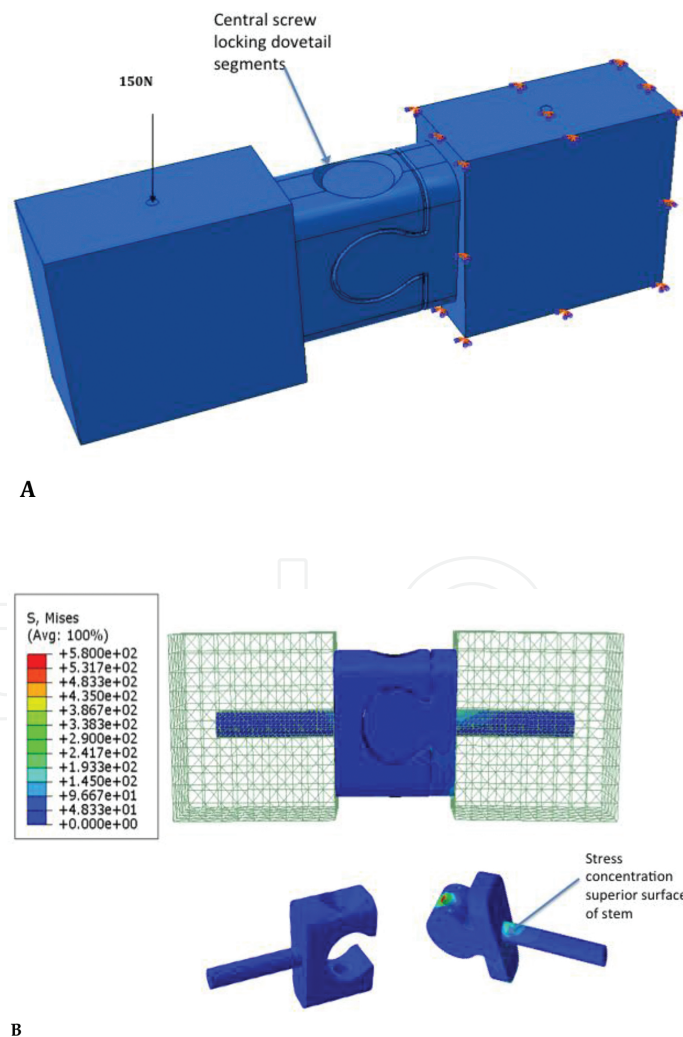


Figure 8. (A) Rendering of endoprosthesis mounted in bone blocks. (B) Von Mises stress distribution.

The finite element analysis of the setup showed areas of high stresses accumulating in the screw hole of the connection screw as well as on the superior surface of the stems. The von Mises stress recorded a maximum of 188.838 MPa, which is way below the strength of titanium alloy at 897 MPa. This corresponds well with the results of the fatigue testing with had failure of crack lines in the same areas. The conclusion of this bench top experiment was that although the forces are way below the material strength of titanium, micro-cracks as well as areas of stress concentrations from the indentation of the screw threads can lead to eventual failure over a long time of loading at a much lower load level. The connection problem of the modules seemed to have been solved.

A human-sized mandible synthetic was scanned with a cone beam CT scan to get the geometrical information of the mandible [31]. A synthetic mandible was used due to biohazard concerns. Due to the nature of the cone beam CT, using the DICOM information without alteration tended to produce a mesh with a lot of sharp, irregular and thin triangles. This was re-meshed with 3-Matics (Materialise, Belgium) into linear tetrahedrons. The mesh was dense enough to justify the use of linear elements to save on computational resources. The teeth were also removed digitally as it does not contribute structurally to the mandible.

The endoprosthesis was modelled as previously described above (termed Case I) and then the stem length was shortened (Case II) followed by Case III where the length of the body was doubled to 30 mm (**Figure 9**).



Figure 9. Endoprosthesis in mandible model.

A standardized defect as well as boreholes was created to fit the dimensions of Case I, II and III on the right side of the mandible using Abaqus. The model was assumed to be made up of only cortical bone, the bone was assumed to be isotropic and the values and vectors of the muscle pull as well as the joint reaction forces were taken from the literature. The boreholes and stems were assumed to be bonded just like the previous FEM study of the experimental

setup. Case I was used as the model for the standard endoprosthesis design, Case II and III were used as models for looking at the effects of a decrease in stem length as well as an increase in defect length, respectively. A 300 N load was applied directly in the incisor region. Several studies from the literature have supported our assumptions to be relatively accurate while saving on computational resources. As a form of validation, a convergence check was conducted on a mandible with quadratic elements, which gave a finer mesh. The differences were less than 10%.

The analysis was conducted and divided into three separate parts:

- 1) Stress in the endoprosthesis
- 2) Stress in the mandible
- 3) Deflection of the mandible

(1) Stress in the endoprosthesis

Under the prescribed loading conditions, the intact mandible bent upwards almost equally on both sides. As this model was not totally symmetrical, there were minute differences on both sides. With a defect in place, reconstructed with an endoprosthesis, the mandible became less stiff, causing the left intact side to arch less than the right reconstructed side. This led to the mandible shifting to the left by 0.354 mm. There was a tendency for the endoprosthesis to bend outwards. There was little difference between Case I and II, which led to the conclusion that a case could be put forward for a shorter stem to reduce the amount of hardware needed. A longer/larger defect, as in Case III led to a tendency for separation at the module connection, leading to a possibility for eventual loosening of the connection screw in the long term. Although there was a tendency for separation at the module connection, the stresses in the connection screw were all below 100 MPa, below the material strength of titanium. The stress distribution for Case I and II was similar, but with slightly larger magnitudes in Case II. In Case III, however, since it was a larger defect, the endoprosthesis tended to bend more at the stem, although, again, the stresses were below the strength of titanium at 898 MPa (**Figures 10–12**).

(2) Stress in the mandible

With regard to the stresses within the mandible, areas of stress concentration were in the left condyle (due to unequal deflection of the mandible from torsional stress), lower border of the mandibular body close to the abutment with the endoprosthesis and top edge of the hole in the stumps. There tended to be a pull out tendency of the stem from the holes, as experienced in the experimental setup. This was worse in Case II and III, with the peak stress exceeding the material strength of cortical bone at 85 MPa. The condyles were restricted from moving in this model, while in real life, there would certainly be movements that could dissipate stress; thus there is some mitigation factor in vivo. Whether this would be true could only be answered in animal experimental models.

(3) Deflection of the mandible

Deflection of the mandible was measured with respect to an x - y - z axis in three dimensions at points of interest. Since the boundary conditions were applied to the incisors and condyles, this caused any deflection to show up in the body of the mandible. The greatest displacement for Case I was 0.638 mm, for Case II 0.8 mm and III 0.608 mm.

The conclusions drawn from the study were that the modular endoprosthesis in its current dimensions should be adequate for small defects. Altering the stem dimensions by shortening it showed a slight increase in magnitude but no significant alteration to the stress distribution. A larger defect, however, would be more difficult to reconstruct and more studies needed to be done.

This work was followed by Pinheiro and Alves [32] who performed a finite element analysis for an endoprosthesis that was not modular and comprised of a solid component, which was customized. This was a feasibility study in which the authors removed the screw stem component and designed a stem that looks to be press fit. The endoprosthesis performed well based on the finite element study. There was no tendency for separation of the module due to the entire prosthesis being a solid framework and the stress distribution as well as the displacement field were very similar to that of the intact mandible and yet did not exceed the material strength of titanium.

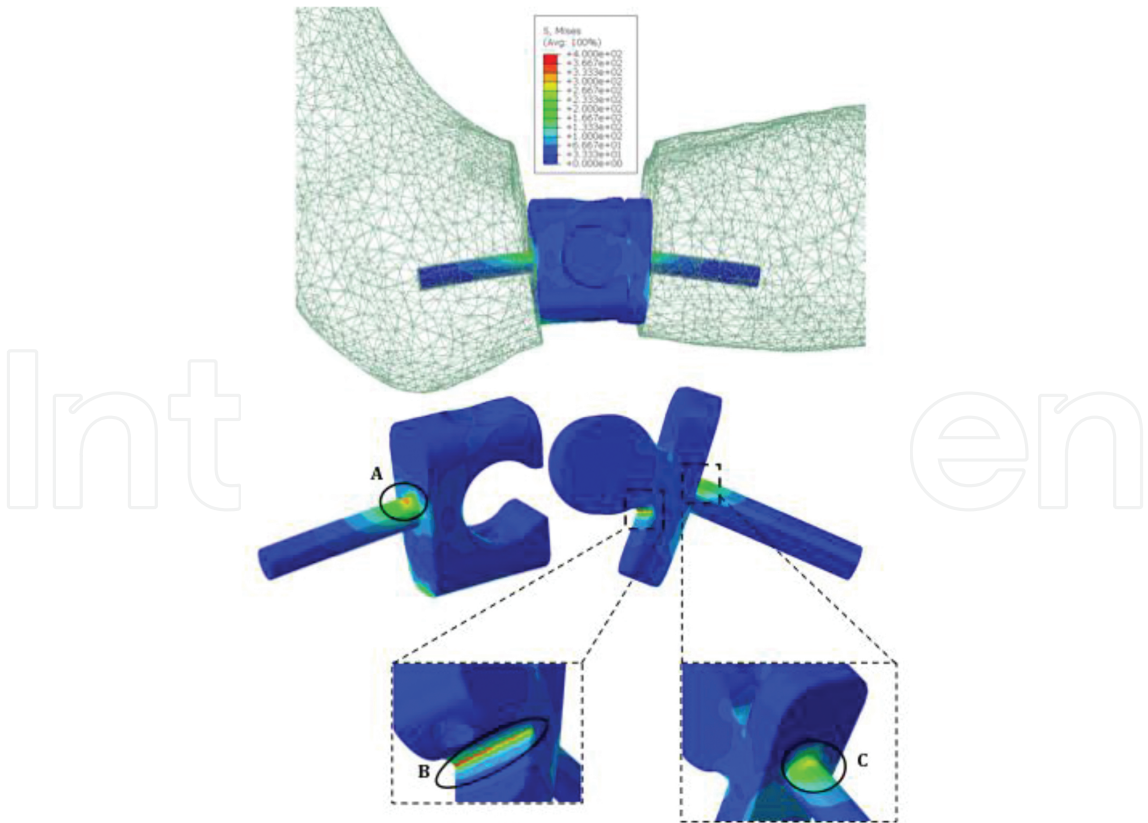


Figure 10. Case I Von Mises stress distribution.

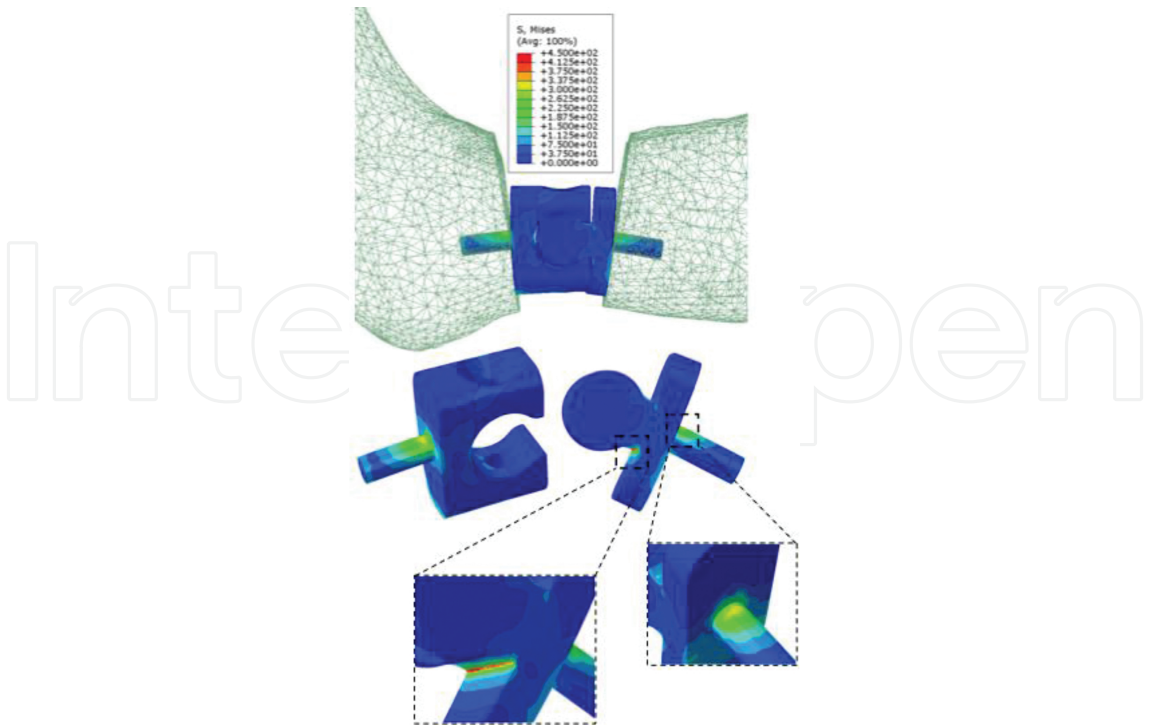


Figure 11. Case II Von Mises stress distribution. The stem length has been halved.

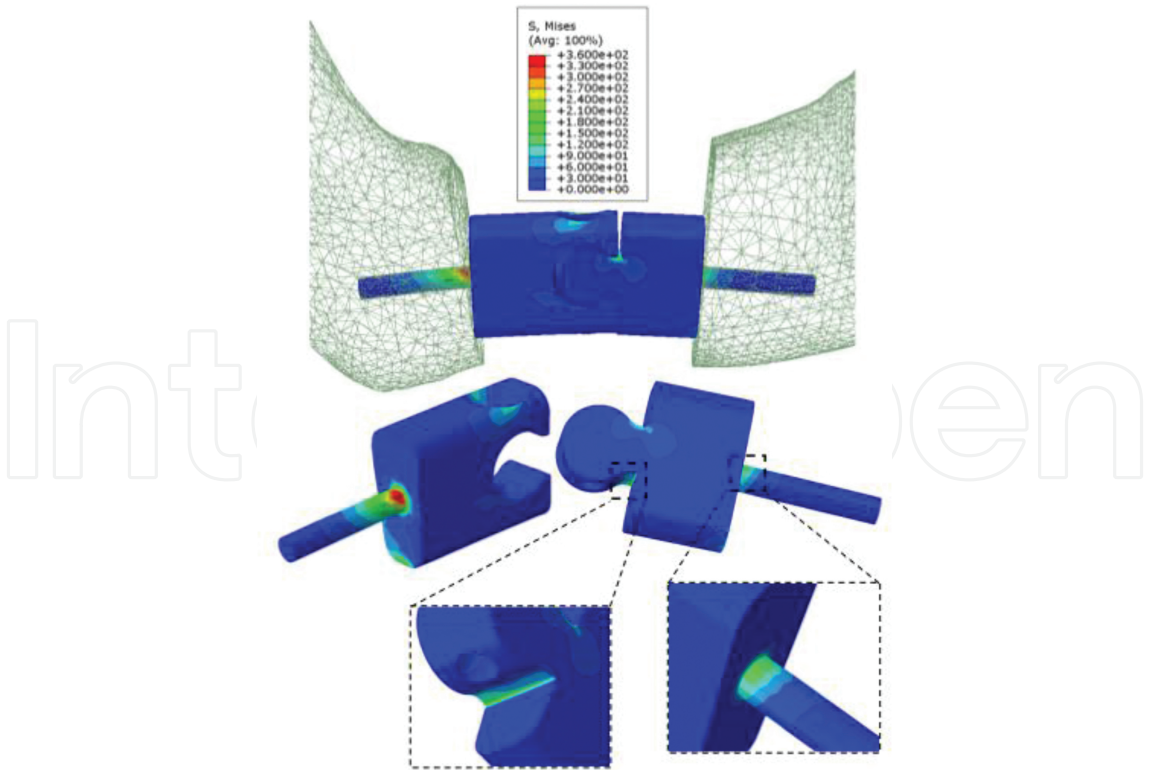


Figure 12. Case III Von Mises stress distribution. The defect length has been doubled.

6. Conclusion

Finite element analysis is a good method to analyse solid mechanics. It has been used extensively in studying the forces in the long limbs as well as the bone plates and the pattern of bone resorption and formation. In the field of head and neck surgery, there are much fewer studies. A number of studies have been recently published as surgeons notice problems with their methods of reconstruction. We have looked at some of the studies and although not exhaustive, it should serve as an illustration to how it is used to look at potential failure of mandibular reconstruction.

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