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Title: Comparison of Three Internal Fixation Techniques in Mandibular Sagittal Split Osteotomy, A Finite Element Analysis

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SUMMARY

The sagittal split osteotomy is a versatile technique used in Mandibular orthognathic surgery. There are many types of internal fixation techniques available to provide fixation following the mandible split. The choice of fixation is mainly on surgeon preference, and can range from bi-cortical screws fixation and miniplate fixation.

This study used Finite Element Analysis to investigate biomechanical properties of three internal fixation techniques. A pre-operative CT scan of a patient's skull was used to develop a half fully dentate mandible computer model. A sagittal split osteotomy was performed on the computer model and all fixation techniques applied to the model. Finite element analysis was used to study the effect of different fixation techniques on various mandibular movements and force magnitudes.

The results of this study have shown consistently that the bi-cortical screws fixation records the least stress and displacement in all simulation models. Stress is mostly concentrated in the inferior-distal screw for the bi-cortical screws fixation whereas the stresses in miniplates are generally dissipated in the connector region of the plate. The 1.7mm miniplate was the least rigid fixation. The stresses in surrounding bone of the fixations were variable for each technique. Miniplate fixations had higher bone stresses in the setback movement but lower with mandibular advancements.

Within the limitations of the study, the application of bi-cortical screws has shown to be the most rigid fixation, although with increased stresses in surrounding bone at mandibular advancements. The 1.7mm and 2.0mm miniplates are less rigid than the bi-cortical screws but handled stresses within the ultimate yield strength.

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1. TERMS AND ABBREVIATIONS

The terms and abbreviations used in this thesis are defined as below for ease of reference.

Term	**Definition**
Anisotropic	A material that consists of more than one property in various areas and direction
Bi-cortical screws	Screws that penetrate both cortices of bone
Biomechanical	Study of the action of external and internal forces on the living body, especially on the skeletal system
CT	Computer tomography - computer-processed combinations of many X-ray measurements taken from different angles to produce cross-sectional images of specific areas of a scanned object
DICOM	Digital imaging and communications in medicine. It is an image format usually containing CT images
Discretization (Meshing)	The process of changing a solid body into smaller finite elements
Elastic Modulus (Young's Modulus)	Measurement of stiffness of a solid material
FEA	Finite Element Analysis
Hooke's Law	The principle of stress imposed on a solid is directly proportional to the strain produced, within the elastic limit
IGES	Initial Graphics Exchange Specification, file format that allows the digital exchange of information among computer-aided design (CAD) systems

Invesalius	Computer software that allows conversion of DICOM images into STL
In-vitro	Studies performed outside of biological context
Isotropic	A material that is uniform in all areas and direction
Mono-cortical screws	Screws that penetrate a single bone cortex
Netfabb	Computer software that allows editing and repair of surface polygons
Newton (N)	A measurement unit for force
NURBS	Non-uniform rational basis spline (NURBS) is a mathematical model commonly used in computer graphics for generating and representing curves and surfaces
Pa (Pascal)	Pascal, SI unit to quantify stress which is equivalent to 1N per square meter. 1 Mega Pascal (MPa) = 1000 Pa, and 1 Giga Pascal (GPa) = 1000 MPa
Poisson's ratio	The ability of a structure to resist deformation in a direction perpendicular to that of the applied load.
Rhinoceros	Computer software that allows creation and manipulation of NURBS
Simulation in Solidworks	An internal programme of Solidworks that runs FEA
Solidworks	Computer software that allows creation, editing and manipulation of 3D solid models
STL	Stereolithography, a file format used to represent surfaces with a series of triangles
Strain	The measurement of the deformation of a material

Stress	The force per unit area applied to an object.
Von Mises Stress	The equivalent tensile stress used to predict the yielding of materials, when they are placed under loads from different directions.

2. INTRODUCTION

Providing adequate stability after orthognathic surgery is important to minimize complications and relapse. Following osteotomy of the mandible, the distal and proximal segments are fixed together with a certain technique. The fixation types used in mandibular orthognathic surgery are usually internal fixation can vary, and over the years have switched from wires, to screws, and now miniplates. It is not standardized and usually depends on the surgeon's preference. The aim of providing stable fixation onto bone segments is to facilitate good bone healing, early mobilization and to prevent relapse.

Internal fixation techniques can be rigid, semi-rigid or non-rigid. Rigid internal fixation can be defined as 'any form of bone fixation in which otherwise deforming biomechanical forces are either countered, or used to advantage to stabilize the fracture fragments, and to permit loading of the bone as to permit active motion'(1). In other words, it is a form of hardware that is applied directly onto bones, and is strong enough to prevent any undesired movements between bone segments. For example, the use of 3 bi-cortical screws in mandibular osteotomies, or large compression plates in the treatment of bone fractures. Edward Ellis explains that non-rigid fixation is any form of bone fixation that is not strong enough to prevent interfragmentary motion across the fracture, when actively using the skeletal structure(2). Earlier, the treatment of mandibular fractures with wires was usually supplemented with intermaxillary fixation, during the healing period, which indicates that the fixation technique was not rigid enough for movements, and only purposed to reduce the fracture.

Historically, wires have been used as a form of fixation in mandibular osteotomies. The first Obwegeser technique in mandibular sagittal split osteotomy used wires through the superior and medial cortices(3). In 1974, Bernd Spiessl introduced the use of the first rigid fixation in orthognathic surgery. He applied compression screws in the mandible. Hans Luhr in 1979 introduced mini bone plates used in midfacial and mandibular orthognathic surgeries(4). The introduction of screws and miniplates in maxillofacial surgery allowed surgeons to provide better treatment. The clear advantages of these fixation techniques are that their application is much faster, easier, and has proven to give better stabilization of bone segments/fragments. Patients heal quicker and have faster recovery as fixation methods get smaller, and postoperative complications lessens.

It is more common to see the use of bi-cortical screws and miniplates in mandibular trauma and orthognathic surgeries today. There is no consensus on which technique should be used, and is usually the surgeon’s preference. A systematic review comparing bi-cortical screws and miniplate fixation following sagittal split osteotomy and mandibular advancement has shown that there is no significant difference in skeletal stability, between the two techniques, although bi-cortical screws have shown to be more stable (5). Removal of hardware following surgery is also not indicated unless symptoms arise such as infection, discomfort, wound dehiscence and palpability. Verweij et al. carried out a systematic review in 2016 and have found that removal of bi-cortical screws are significantly less than titanium miniplates (6). The reasons behind this may be due to higher surface area and ‘foreign body reaction’ of the miniplates, which can increase susceptibility to infection (7).

Many studies have investigated the biomechanical properties of different internal fixation techniques used in mandibular orthognathic surgeries. These include in-vitro, and finite element analysis studies. There is a growing interest in the use of finite element analysis (FEA) in the studies of human oral and maxillofacial biomechanics including maxillofacial trauma, zygomatic inplants, dental implantology and orthognathic surgery (8-12). The finite element method is a numerical method for solving problems of engineering, and mathematical physics. By using an FEA software, the biomechanical properties such as displacement, stress, and strain of bone and fixation techniques can be analysed in various situations and loads.

A smaller miniplate than 2.0mm is not commonly used in mandibular fixation. Smaller miniplates may allow less foreign body reaction, palpability, and other complications, while providing adequate strength for fixation. As to date, no study has included a much smaller miniplate than 2.0mm in mandibular orthognathic surgery. This study investigated the biomechanical properties of three internal fixation techniques (2.0mm bi-cortical screws, 2.0mm miniplate with mono-cortical screws and 1.7mm miniplate with mono-cortical screws) used in mandibular orthognathic surgery, subjected to different biting forces and movements (advancements and setback).

3. AIM AND OBJECTIVES

3.1. AIM

To compare the biomechanical behaviour of three different internal fixation techniques in mandibular sagittal split osteotomy.

3.2. OBJECTIVES

1. To create a computer model of a dentate human mandible.
2. To determine the best mesh setting, and simulation model for Finite Element Analysis.
3. To analyse the displacement and stresses transferred to fixations, and surrounding bone during various loading conditions in mandibular setback movement.
4. To analyse the displacement and stresses transferred to fixations, and surrounding bone during various loading conditions in mandibular advancement movements.

3.3 HYPOTHESIS

1. The 1.7mm miniplate fixation will have similar stresses in fixation compared to the 2.0mm miniplate and 2.0mm bi-cortical screws.
2. The 1.7mm miniplate will show similar stresses in bone surrounding the fixation compared to 2.0mm miniplate and 2.0mm bi-cortical screws.
3. The 1.7mm miniplate fixation will have similar stability compared to the 2.0mm miniplate and 2.0mm bi-cortical screws.

4. LITERATURE REVIEW

4.1 Bilateral Sagittal Split Osteotomy

Orthognathic surgery in the mandible is a common procedure in correcting dentofacial abnormalities. Bilateral sagittal split osteotomy (BSSO) as shown in figure 3.1, is one of the most common surgical procedure used in mandibular orthognathic surgery. A sagittal split is made in the ramus which allows various movements of the split bone segments including advancement, reduction or setback, and rotation.

E.Steinhauser in 1996 described the historical development of mandibular orthognathic surgery in his article very clearly. R.Trauner described various approaches to the mandibular osteotomies such as the inverted-L osteotomy for prognathism correction in 1955. He was also known to have trained Hanz Kole and Hugo Obwegeser in orthognathic surgery (4). Obwegeser published the world-reknown intra-oral sagittal split osteotomy of the mandible in 1955 (3). The technique involves 2 parallel cuts made medially above the lingula and laterally just above the angle. An osteotome is then used carefully to split the cortices. This produces an area of bone overlap that can be used for mandibular advancements and setbacks. DalPont in 1961, an Italian surgeon made a modification to the original Obwegeser design by extending the lateral cortex cut to the buccal aspect of the second molar(13). This increased the surface area between the two osteotomised segments. Hunsuck in 1968 modified the DalPont design by placing the lingual cut just behind the lingula as he thought that a natural fracture pattern will occur naturally with use of a chisel (14). In 1977, Epker published an article that refined the DalPont buccal corticotomy, where he describes that the buccal cut should extend into the inferior border of the mandible in order to prevent ‘bad splits’ (15).

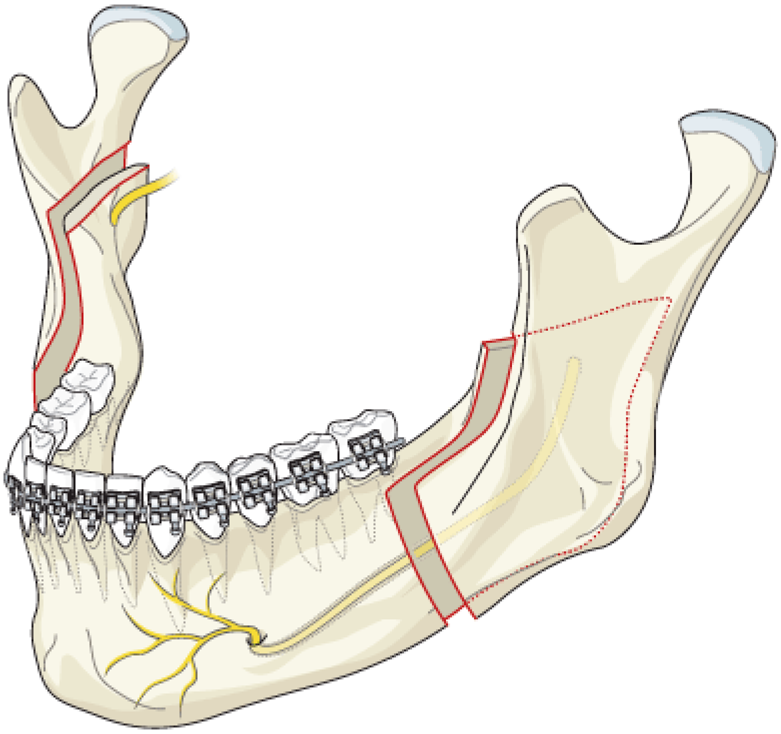


Figure 4.1 Bilateral sagittal split osteotomy of the mandible

4.2 Internal Fixations in Orthognathic Surgery

4.2.1 History

The mandibular sagittal split osteotomy has become the most used technique due to its versatility in providing various movements, including advancements, setbacks and rotations. Following the split and planned movement of the mandible, the proximal and distal segments are fixed together with internal fixation. Most commonly used today are bi-cortical screws or miniplates, rather than wires.

Internal fixation with wires were used widely in the treatment of bone fractures including maxillofacial trauma. It was also the choice of fixation in orthognathic surgery prior to the introduction of rigid internal fixation such as screws by Spiessl in 1974 and miniplates by Luhr in 1979(4). At the time, fixing the separated bone segments ‘rigidly’ was crucial for success, as according to one of the Swiss Association for the Study of Internal Fixation (AO/ASIF) four biomechanical principles in fracture management (16). This requires larger size of fixation, more hardware insertion, and utilization of a compression type of fixation. Interestingly in 1979, Martin Allgower viewed internal fixation differently. As quoted in his article Internal Fixation of Fractures: Evolution of Concepts, ‘Too much fixation can result in cancelization of cortical bone under a plate by relieving nearly all the physiological stimulus of weight-bearing, and movement, whereas too little fixation leads to micromotion and resorption’(1). This understanding may have given rise to a new term of internal fixation which is ‘semi-rigid or functionally stable’ fixation.

Internal fixation in the maxillofacial region is not always rigid. The use of miniplates in the treatment of mandibular angle fractures as shown by Champy in 1978, allows the fixation to stabilize the fracture, although some interfragmentary motion is possible (17). Thus has been proven successful clinically, and is shown by a systematic review and meta-analysis of 20 studies in 2014 investigating the best method of fixation in mandibular angle fractures with the lowest rate of postoperative complications (18). The study concluded that the use of a single miniplate in the superior or lateral border of the mandible decreased the risk of postoperative complications by 37% compared to 2 miniplates. Champy’s technique takes advantage of the reduced bony fragments, which is supported with fixation stabilization (17). Ellis in 2004 advices that fixation placed over a fracture or osteotomy gap must provide sufficient strength to allow transmission of functional forces across the gap without alteration of the occlusion(19). He also suggests that a very thin bone plating system can provide stability in most fractures and osteotomies. Thus, smaller fixation such as 2.0mm bi-cortical screws and miniplates are more commonly used in orthognathic surgery as they provide the necessary stability, less hardware, less surgical exposure, as well as being easier and faster to apply.

Internal fixation implies a fixation technique that is applied directly to bone. In mandibular osteotomies the use of internal fixations has avoided or decreased the lengthy period of intermaxillary fixation, allowed early mobilization of the mandible, and may have greater patient acceptance (4). The commonly used techniques for internal fixation for mandibular osteotomies are bi-cortical screws or miniplates with mono-cortical screws. These techniques come in various sizes and lengths according to the manufacturers.

4.2.2 Bi-cortical screws fixation

This technique was first introduced by Spiessl in 1974(4) .The name of the fixation itself implies that the fixation incorporates into both cortices of the bone. This makes them much longer than mono-cortical screws. The range of length is usually between 10-15mm. Bi-cortical screws can either be positional or compressive (lag screws)(19). A guiding hole is made through both cortices of the bone and the screw inserted perpendicular to the fracture line. There are several ways and permutations of the bi-cortical screws in mandibular orthognathic surgery including the use of 2 superior border screws, 3 superior border screws, 3 screws in a triangular pattern and others. There is no consensus on which permutation or number of screws required and usually is surgeon preference. Although, some in-vitro and FEA studies have shown that the use of 3 bi-cortical screws in an inverted – L pattern is the more stable approach (20-26).

Bi-cortical screws offer greater rigidity and stabilization in mandibular osteotomies as shown in various in-vitro and in vivo studies (21, 25-29). There are some known complications reported in the literature with the use of bi-cortical screw fixations. Clinically, the application of bi-cortical screws requires a trans-buccal approach which results in a small skin scar, although less noticeable in the Caucasian population, it might result in keloid formation for

African and Asian population (30). This technique may also cause increased torque in the condylar region which may lead to condylar resorption and temporomandibular symptoms (31). Bouwman et al. in 1995 have reported increased neurosensory disturbances following bi-cortical screws placement, possibly due to damage to the inferior alveolar nerve (30).

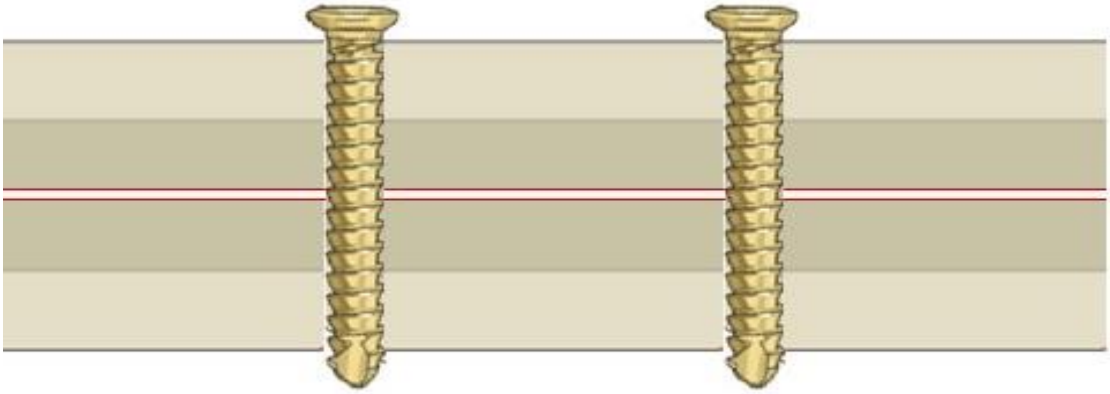


Figure 4.2 Bi-cortical screw fixation

4.2.3 Miniplate fixation

Miniplate fixation allows the use of shorter, usually mono-cortical screws which ranges from 5 to 8mm in length. While the screws anchor onto the bone cortex, the plate acts as the bridge connecting the two bone segments. The miniplate must be well adapted to the bone surface. Numerous sizes of miniplates are available on the market and can range from 1.2mm to 2.4mm in screw hole diameter for use in the maxillofacial region. They also have variable designs including continuous holes, miniplates with separator gaps of various length, locking and non-locking, and sliding miniplates on the superior border (32-34).

Biomechanical studies have shown that miniplates are generally less rigid than bi-cortical screws, although they provide similar skeletal stability in the long-term (5). Stoelinga and Bortslap in 2003 explain the advantages of miniplate fixation includes; 1) no stab incision is required extra-orally which imposes risks of scarring and injury to the facial nerve, 2) malleability of titanium plates allows better adaptation to the proximal and distal bone segments, and reduces condyle torque, 3) mono-cortical screws can avoid damage to the inferior dental nerve, and 4) when necessary, removal of miniplates is easy and can be done under local anaesthesia (35). Complications of miniplates was explained by Ochs in 2003, whom claimed that titanium miniplates can cause chronic irritation and palpability underneath the gums (36). The increased surface area may also cause more foreign body reaction and infections (7). Loukota and Shelton in 1995 have shown that the diverse designs in miniplates do affect their compression and tensile strengths, and while they all perform beyond requirements of clinical situations, the failure in miniplate fixation is mostly caused at the bone-screw interface (32).

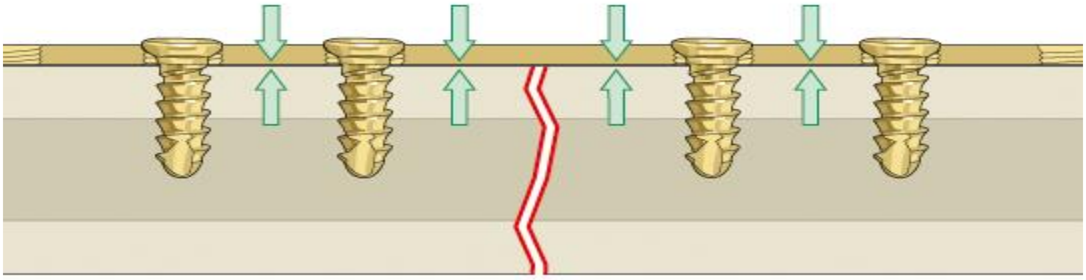


Figure 4.3 Miniplate with mono-cortical screw fixation

4.3 Biomechanical studies of various fixations in mandibular orthognathic surgery

A literature search of biomechanical studies comparing various fixation techniques in mandibular osteotomies was carried out. An electronic database search was used without restriction concerning to date of publication: using PubMed and MEDLINE. The search strategy keywords were mandible osteotomy, orthognathic surgery, internal fixation, biomechanical and finite element analysis.

The inclusion criteria were: studies involving mandibular osteotomies, studies using internal fixation which includes titanium screws and plates, and computational studies involving finite element analysis (FEA). Exclusion criteria includes; animal studies, other anatomical site, trauma or fractures, using non-biomechanical methods, use of biodegradable plates/screws, language other than English.

22 articles published between 1995 and 2016 met the eligibility criteria. Out of the 21 articles, 9 articles involving in-vitro non-computational studies and 13 articles involving computational studies using finite element analysis (FEA). Table 4.1 shows a description of selected relevant studies.

Table 4.1 Biomechanical studies of fixation techniques in mandible osteotomy

Author	Year	Country	Design
Albougha et al. (20)	2015	Syria	Finite element analysis
Bohluli et al. (21)	2010	Iran	Finite element analysis
Brasileiro et al. (26)	2009	Brazil	In-vitro study
Brasileiro et al. (25)	2012	Brazil	In-vitro study

Author	Year	Country	Design
Chuong et al. (37)	2005	USA	Finite element analysis
Erkmen et al. (22)	2005	Turkey	Finite element analysis
Erkmen et al. (23)	2005	Turkey	Finite element analysis
Hammer et al. (38)	1995	Switzerland	In-vitro study
Lee Ming Yih et al. (24)	2010	Taiwan	Finite element analysis
Maurer et al. (39)	1999	Germany	Finite element analysis
Maurer et al. (40)	2003	Germany	Finite element analysis
Oguz et al. (27)	2015	Turkey	In-vitro study
Oguz et al. (41)	2009	Turkey	Finite element analysis
Pereira Filho et al. (42)	2010	Brazil	In-vitro study
Peterson et al. (29)	2005	USA	In-vitro study
Ribeiro et al. (2010) (43)	2010	Brazil	In-vitro study
Ribeiro et al. (2012) (44)	2012	Brazil	In-vitro study
Sato et al. (45)	2012	Brazil	Finite element analysis
Scaf de Molon et al. (46)	2011	Brazil	In-vitro study
Sindel et al. (28)	2014	Turkey	Finite element analysis
Stringhini et al. (47)	2016	Brazil	Finite element analysis
Takahashi et al. (48)	2010	Japan	Finite element analysis

4.3.1 In-vitro studies

The 9 in-vitro studies used polyurethane models rigidly held onto a steel arm with a steel rod applied vertically onto the occlusal surface. The comparison of fixation stability is translated into how much force is required to displace the distal segment of the split mandible. Out of the 9 studies, 7 of them compared bi-cortical screws, and miniplate with mono-cortical screws fixation. One study compared 2 different diameters of bi-cortical screw fixation on a mandible osteotomy(46), whereas another compared only different types of miniplates(44). The fixation methods in in-vitro studies varied from bi-cortical screws in inverted-L pattern, 4-hole to 6-hole miniplates with mono-cortical screws, locking miniplates, and sagittal miniplates.

Two studies applied vertical forces from the incisors(46, 49). Two other studies applied vertical forces from premolars 14, 15 while another two from Ribeiro et al. applied vertical forces to molar region (43, 44). Both studies by Brasileiro et al. applied forces to incisors and added another force to the lateral buccal of molars (25, 26). The study by Petersons et al. compared forces from both incisor and molar loadings (29). Forces were applied using servo-hydraulic material testing unit.

Osteotomy cuts involved in the polyurethane mandible are all sagittal split osteotomies which only 3 studies specifically mentioned the name of cut; Obwegeser and Obwegeser Dal Pont(27, 41, 49). 7 studies opted for advancements after osteotomy with 2 studies applied 4mm advancement(43, 44), 4 studies with 5mm advancement(26, 27, 42, 46) and one applying 7 mm advancement(29). Hammer et al.(49) did not apply any movements and Brasileiro et al.(25) opted for 5mm setback.

All works that used bi-cortical screws in an inverted-L pattern showed the best stability, and less displacement when compared to other fixations. Hybrid fixation of miniplates with an added bi-cortical/positional screw showed results that are superior than miniplates alone, but inferior to the bi-cortical screws.

4.3.2 Computational studies

13 studies were identified using computational studies with finite element analysis (FEA) to analyse the biomechanical properties of different fixation systems in mandibular osteotomies. A summary of the FEA studies are shown in Table 4.2.

Table 4.2 Biomechanical studies of mandibular osteotomy using the Finite Element Analysis

AUTHOR	FEA SOFTWARE	MANDIBLE DESIGN	FORCE	FIXATION	MODEL MESH	MATERIAL PROPERTIES	RESULTS
Albougha et al. 2015	ANSYS v. 13 (ANSYS, Inc., Canonsburg, PA, USA)	Hemi-Mandible, 2 layers, fixed on condyle and lateral ramus (masseter). Sagittal Split, did not mention movement	500N to posterior teeth	bicortical screws; • straight line with 60 degrees angulation • straight line with 90 degrees angulation • L-pattern • inverted-L pattern 1 x 4-hole miniplate 2 x 4-hole miniplate T-shaped miniplate Y-shaped miniplate 1 x 4-hole miniplate + 1 bicortical screw	Not mentioned	Cortical: 14GPa / *P: 0.3* Cancellous: 1.5GPa / P:0.3 Titanium: 114GPa / P: 0.34	Straight miniplate most favourable
Bohluli et al. 2010	ANSYS software (v 10; ANSYS, Canonsburg, PA)	Hemi-Mandible, single bone layer, fixed on condyle. Sagittal Split, did not mention movement	75, 135 and 600 N to posterior teeth	1 bicortical screw 2 x bicortical screw 3 x bicortical screw, L pattern 3 x bicortical screw, inverted-L pattern 1 x 2-hole miniplate 1 x 4-hole miniplate 2 x 4-hole miniplate 1 x 4-hole square plate	0.9mm^3 cube meshing	Bone: 14.8GPa / P: 0.03 Titanium: 114GPa / P: 0.33	3 x bicortical screws in inverted-L pattern shows best stability
Chuong et al. 2005	ANSYS 6.1 software (ANSYS Inc, Canonsburg, PA)	Full mandible with no teeth, isotropic, fixed on condyle. Bilateral Sagittal Split, Advanced 8mm	0.5 - 1kg to anterior segment	3 x bicortical screws 1 x 4-hole curved miniplate	Not mentioned	Bone: 13.7GPa / P: 0.3 Titanium: 120GPa / P: 0.36	3 x bicortical screws in inverted-L pattern shows best stability
Erkmen et al. (advancement) 2005	MSC MARC Software (MSC Corporation, Santa Ana, CA 92707, USA)	Hemi-Mandible, 2 layers, fixed on condyle. Sagittal Split, Advanced 5mm	500N to posterior teeth	3 x bicortical screws, inverted-L patterm 2 x bicortical screws, vertical 2 x 4-hole miniplate 1 x 4-hole miniplate	122717 elements, 25048 nodes	Cortical: 14.8GPa / *P: 0.3* Cancellous: 1.85GPa / P:0.3 Titanium: 105GPa / P: 0.33	3 x bicortical screws in inverted-L pattern shows best stability
Erkmen et al. (setback) 2005	MSC MARC Software (MSC Corporation, Santa Ana, CA 92707, USA)	Hemi-Mandible, 2 layers, fixed on condyle. Sagittal Split, Setback 5mm	600N to posterior teeth	3 x bicortical screws, inverted-L patterm 2 x bicortical screws, vertical 2 x 4-hole miniplate 1 x 4-hole miniplate	122717 elements, 25048 nodes	Cortical: 14.8GPa / *P: 0.224* Cancellous: 1.85GPa / P:0.3 Titanium: 105GPa / P: 0.33	3 x bicortical screws in inverted-L pattern shows best stability
Lee Ming Yih et al. 2010	(ANSYS, v 8.0, Swanson Analysis Inc., Houston, PA, USA)	Full mandible with no teeth, isotropic, fixed on condyle. Sagittal split Setback 10mm	150N to anterior	2 x bicortical screws 3 x bicortical screws (5 various configurations)	353626 elements, 67837 nodes	Cortical: 13.3GPa / *P: 0.224* Cancellous: 1.33GPa / P:0.224 Titanium: 117GPa / P: 0.33	3 x bicortical screws in inverted-L pattern shows best stability

AUTHOR	FEA SOFTWARE	MANDIBLE DESIGN	FORCE	FIXATION	MODEL MESH	MATERIAL PROPERTIES	RESULTS
Maurer et al. (1999)	ANSYS	Hemi-Mandible, single bone layer, fixed on condyle. Sagittal Split	170N to posterior teeth	3 x bicortical screws (1.5mm vs 2.0 mm diameter)	19845 elements, 4285 nodes	Bone: 200MPa Titanium: 610MPa	No significant difference between different screw diameters
Maurer et al. (2003)	ANSYS	Hemi-Mandible, single bone layer, fixed on condyle. Sagittal Split, Did not mention movement	170N to posterior teeth	3 x bicortical screws, inverted-L pattern 1 x 4-hole miniplate	19845 elements, 4285 nodes	Bone: 500MPa Titanium: 610MPa	3 x bicortical screws in inverted-L pattern shows best stability
Oguz et al. 2009	Not mentioned	Hemi-Mandible, 2 layers, fixed on condyle. Sagittal Split, Advanced 5mm	200N to angle of mandible	1 x 4-hole miniplate 1 x 4-hole locking miniplate	153320 elements, 35570 nodes	Cortical: 14.8GPa / *P: 0.3* Cancellous: 1.85GPa / P:0.3 Titanium: 105GPa / P: 0.34	Locking miniplate spreads forces more widely than conventional
Sato et al. 2012	Femap version 10.1 (Siemens PLM Software Inc., Plano, TX)	Hemi-Mandible, single bone layer, fixed on condyle. Sagittal Split, Avanced 5mm	Load until 3mm displacement	3 x bicortical screws, inverted-L pattern 3 x bicortical screws, linear 90 degrees angulation 3 x bicortical screws, linear 60 degrees angulation 1 x 4-hole miniplate 1 x 4-hole locking miniplate	Not mentioned	Bone: 624.4 MPa / P: 0.2817 Titanium: 116GPa / P: 0.34	3 x bicortical screws in inverted-L pattern shows best stability
Sindel et al. 2014	Not mentioned	Full mandible with teeth, 2 layers, fixed on condyle and posterior mandible. Bilateral Sagittal Split Advancement and Setback 7mm	200N to posterior teeth and muscle forces	3 x bicortical screws L pattern 3 x bicortical screws inverted-L pattern 3 x bicortical screws linear (1.5mm vs 2.0mm diameter screws)	Not mentioned	Not mentioned	3 x bicortical screws in inverted-L pattern shows best stability
Stringhini et al. 2016	Ansys Workbench 14.0	Full mandible with 2 bone layers, enamel, dentin, pulp, pdl, fixed on condyle and incisor Bilateral Sagittal Split, Advanced 5mm	Muscle insertions; (temporalis - 235 N, medial pterygoid muscle - 145 N, masseter muscle- 151 N	4-hole conventional 2.0-mm miniplate 4-hole locking 2.0-mm miniplate 6-hole conventional 2.0-mm miniplate 6-hole locking 2.0-mm miniplate 2.0-mm conventional screw 6 mm long 2.0-mm locking screw 6 mm long 2.0-mm conventional screw 15 mm long.	2500662 nodes, 1489170 elements	Not mentioned	Locking miniplate/screw is more stable
Takahashi et al. 2010	SolidWorks2008 (SolidWorks Japan, Tokyo, Japan)	Full mandible, single bone layer, fixed on condyle. Bilateral Sagittal Split Advanced 5mm	260.8N on Molars, 66.7N on incisor	1 x 4-hole miniplate, different orientation	134836 elements, 29582 nodes	Bone: 13.7GPa / P: 0.3 Titanium: 110GPa / P: 0.34	Ob-DalPont osteotomy design provides more stability

In FEA, the mandible model is developed using computer tomography (CT) of a patient's mandible, cadaver or polyurethane synthetic mandible. The sagittal split is carried out virtually and the mandible is advanced, or set back. Fixation techniques are developed according to manufacturer design and applied onto the mandible model. Boundaries and restraints of the mandible model is defined, usually located in the condylar region. Virtual forces are applied either onto the occlusal surfaces, or as muscle forces. Values of material properties of bone, and fixation, are inserted in the FEA software. The assembly of mandible and fixation undergoes meshing or discretization. This is where the model is broken into smaller finite elements.

The FEA software calculates and analyses the displacement in the fixation and bone when force is applied. This then allows calculation of the strain and further on the stress using the Hooke's law and known elastic modulus of the material. The results are shown as coloured diagrams with relevant measurements/units. Von Mises Stress is usually used to measure stresses in ductile materials. It is the equivalent stress used to predict the yielding of materials, when they are placed under different loading conditions.

All the mandibles in FEA underwent a sagittal split osteotomy (SSO) with 6 studies specifically mentioning the Obwegeser-Dalpont osteotomy design. Sato et al. in 2012 opted for the Epker modification of Obwegeser's osteotomy while Takahashi et al. in 2010 compared 3 different modification of the SSO namely; Obwegeser, Obwegeser-Trauner, and Obwegeser - Dalpont.

Out of the 13 computational studies, 7 studies showed superior results in bi-cortical screws, as a fixation in mandibular osteotomies, when compared to miniplates. Similar to in-vitro studies, the bi-cortical screws in an inverted L pattern have been shown to be the most rigid fixation. In studies that did a comparison among miniplates, it has been shown that locking miniplates are able to spread force more widely in the fixation compared to conventional miniplates, and that miniplates placed in the line of Champy show more stability. (20, 27, 47, 48).

The fixation systems reported in the literature include bi-cortical screws and miniplates. Bi-cortical screws in an inverted L pattern have shown to provide the best stability in mandible osteotomies. The use of miniplates in mandible osteotomies have shown to be less stable than bi-cortical screws. Champy et al in 1978 explains that when placed in the most bio-mechanically advantageous area, a small plate can neutralize the functional forces, and permit active use of the mandible during the healing process(17). Although Champy's technique is functionally stable, interfragmentary motion probably occurs to some extent during function and is therefore not rigid fixation. Generally, 2.0mm systems are used in mandibular osteotomies with 4 screws, 2 screws on either side of the osteotomy.

Comparing the amount of "stresses" in fixation and surrounding bone between the FEA studies has proven to be difficult. This is because of the non-standardized values of material properties, fixation techniques, type of mandible used, location of restraints, and bite force loading. Furthermore, the mesh of each mandible model differs in each study. Earlier studies such as Maurer et al. in 1999 recorded only 19845 elements in their hemi-mandible model whereas a more recent study by Stringhini et al. in 2016 recorded 1489170 elements in a full mandible model (39, 47). This is related to the complexity of the model, where the latter included various structures to the model including individual teeth, periodontal ligaments and cortical and cancellous bone layers. Another reason for these differences could also be due to technology advancement and better available computer resources.

In in-vitro studies, synthetic jaw models made from polyurethane were used because they are easy to obtain, inexpensive, and allowed standardization. The main disadvantage of this model is that it has different elasticity values from that of fresh bone, and thus is unable to provide accurate results in the tests carried out. Baccarin et al. in 2011 analysed 3 different synthetic materials used in fabrication of the mandible models for in-vitro testing, and have found that there were significant differences in the results (50).

In contrast, the FEA technique allows the analyses of displacement and the forces required to cause failure of any material used. As the whole problem domain is subdivided into smaller finite elements, stresses in the fixation, and surrounding bone can be analysed. The advantages of doing so includes accurate representation of complex geometry, inclusion of dissimilar material properties, easy representation of the total solution, and capture of local effects(51). However, the technicalities involved to set up this type of study may be a drawback for clinicians, and a collaboration with an engineering or biomechanical department is advised.

There are limitations in the FEA studies especially in the construction of mandible model and the fixation techniques. All FEA studies did not incorporate the inferior dental canal which can affect the sagittal split and placement of the fixation techniques. The condyles are usually fixed and prevents physiological movements and torque when the mandible segments are fixed. Furthermore, majority of the FEA studies simplified the screw-bone interface by applying cylindrical screws and 'bonding' them onto the bone to simulate rigid fixation (20-23, 37, 39, 40).

4.4 Finite Element Analysis

4.4.1 Theoretical Methods

Daryl L. Logan 2011 lists the step by step theoretical methods involved in FEA (52):

1. Discretization (meshing) – producing a solid body of smaller finite elements
2. Define displacement function - This step involves selecting a displacement function of each element. Commonly used functions are linear, quadratic and cubic polynomials.
3. Define the stress/strain relationships
4. Derive the element stiffness matrix and equations
5. Assemble the element equations to obtain the global or total equations and introduce boundary conditions
6. Solve for the unknown degrees of freedom (or generalized displacements)
7. Solve for the element strains and stresses
8. Interpretation of results

4.4.1.1 Discretization

Discretization involves dividing the body into smaller finite elements with associated nodes (points where elements meet) as shown in Fig 4.4. The type and shape of elements depends on the geometry of the body and complexity of the problem that needs to be solved. Elements can exist in 1-dimensional (simple line), 2-dimensional (triangles) and 3-dimensional forms (tetrahedral, hexahedral). The number and size of elements used dictates the accuracy of the results. However, more elements mean a more computational effort and longer solving time. To determine the least number of elements required for more accurate result, a series of simulations were carried out to show evidence of convergence between a numbers of mesh densities for comparison (10). This is called the convergence test and is illustrated by the graph below (Fig 4.5). At point 1, when the mesh is coarse, and number of elements are low, the result of stress value is also lower. However, at point 2 where the mesh is much more refined, and elements increases, the stress value is higher and can differ significantly from point 1. As the mesh undergoes further refinement in point 3, the stress value starts to reach a plateau, and this is where it is said that the mesh has reached convergence. It is acceptable to use the number of elements where the difference of stress value is less than 5% when compared to a finer mesh of more elements.

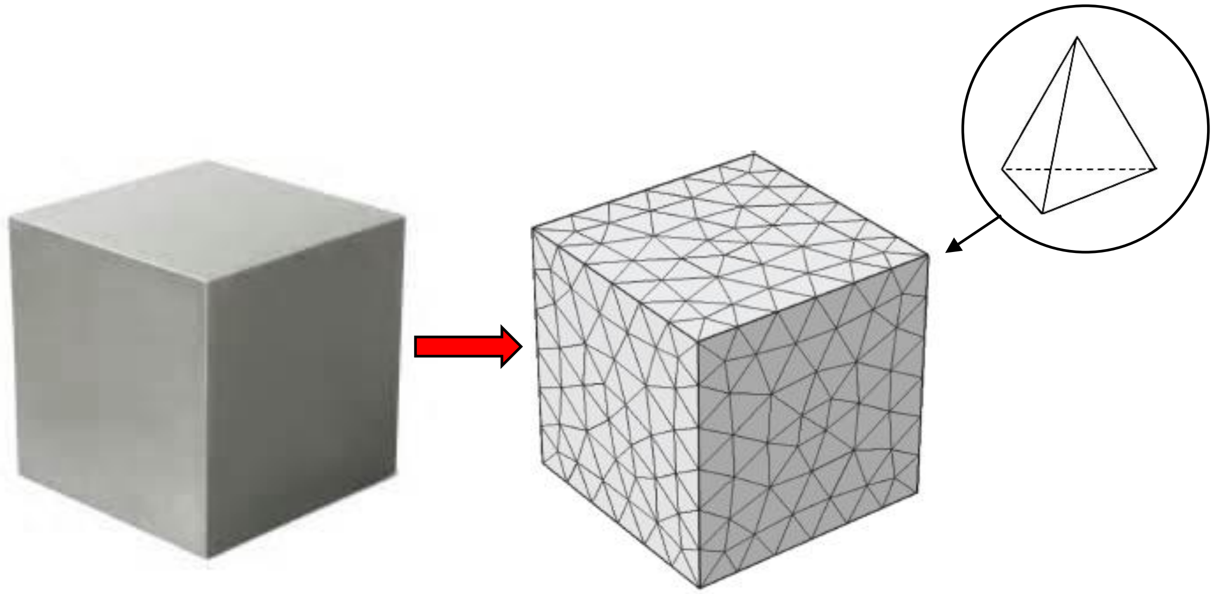


Figure 4.4 Solid cube undergoes discretization which consists of 3-dimensional tetrahedral elements as shown in the circle

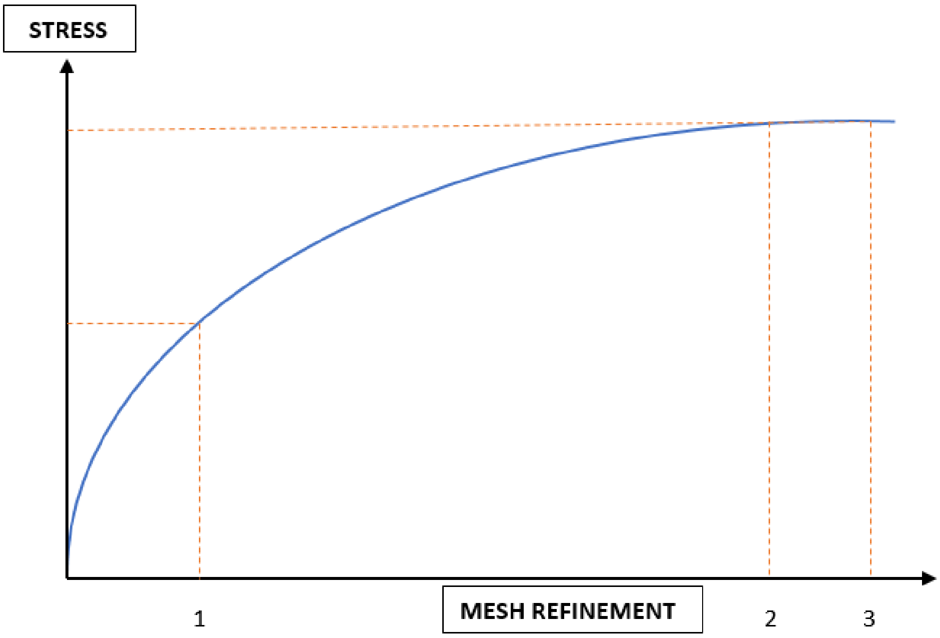


Figure 4.5 Mesh Convergence Graph

4.4.1.2 Material Properties

Defining the properties of the materials used in the FEA study is important, as it is required for the formulation of stress / strain relationship in the elements. The elastic modulus measures an object's resistance to non-permanent deformity. It measures the stiffness of the object. It is defined in Hooke's Law formula as $E = \sigma/\varepsilon$, where E is elastic modulus, σ is stress and ε is strain. The graph (Fig 4.6) illustrates the stress / strain curve for a steel material. Beyond the yield point, the material undergoes plastic deformation.

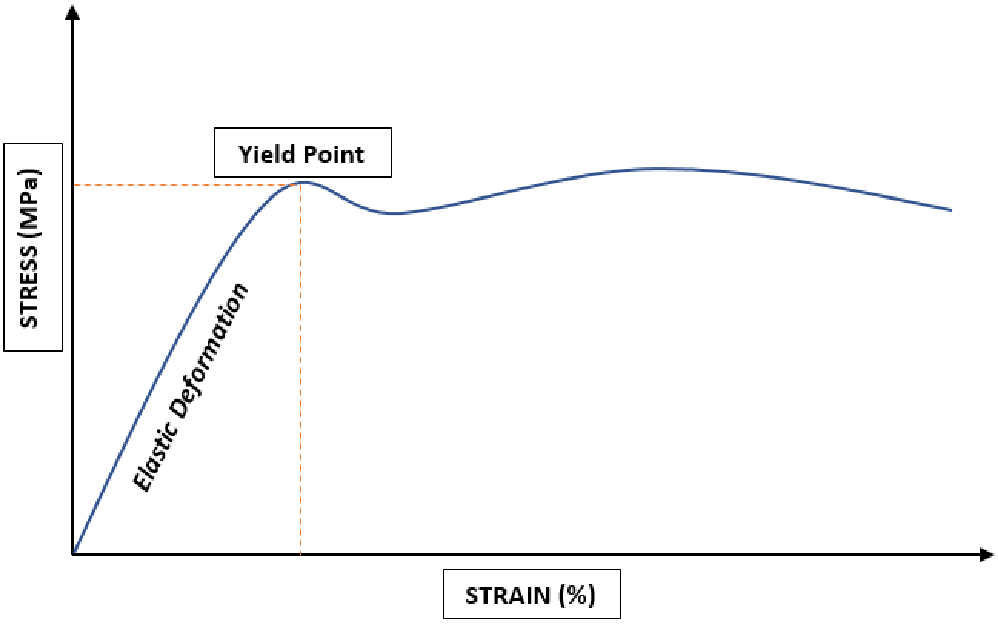


Figure 4.6 Stress - Strain Graph for Steel

Poisson's ratio is the ratio of deformation in the direction perpendicular to the force applied. For example, a rubber band becomes thinner as it is pulled from each side. This can be further explained in Fig. 4.7.

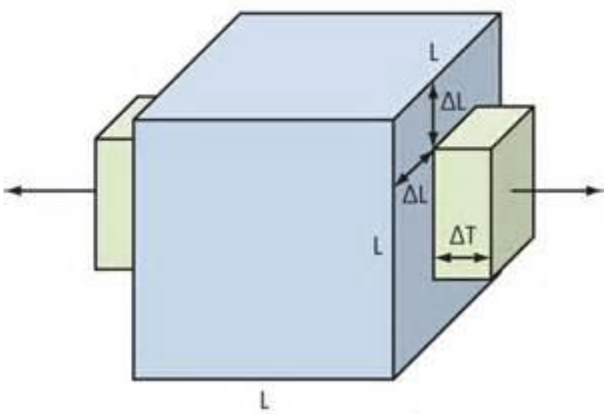


Figure 4.7 Diagram illustrating Poisson's Ratio

Material properties of non-biological origin are readily available, and usually consistent as they have been studied repetitively. For example, titanium, steel, and other alloys. However, the same cannot be said for organic materials, like bone. Bone is a living tissue that is anisotropic and non-homogenous in nature. It consists of a dense cortical and a porous cancellous layer. The microstructure of bone runs in various directions. This makes the modelling of proper bone in FEA difficult, and most studies resolve this with simplification by assuming homogeneity, and isotropy in bone. Strait et al in 2005 studied the effects of various bone isotropy and homogeneity in a monkey (*Macaca fascicularis)* skull model (53). The authors conclude that the precision in the model is highly dependent on the type of study carried out. If the study is to assess gross pattern deformations, then a less detailed model is acceptable. However, when data extracted is required for quantitative analysis, a more precise model is needed to reflect reality.

Previous FEA studies have shown to prefer different values for bone properties, as seen in Table 4.2. These different values in bone properties can affect the results in fixation stability. Logically, bone with a higher elastic modulus would be able to withstand greater forces before undergoing deformation. Erkmen et al. in 2005(22) and Ming Yih et al. in 2010 (24) did not specify their source of data, while Oguz et al. in 2009(41) opted for values used in the former study. Dechow et al in 1993 (54) compared the elastic properties of bone from the supraorbital and the mandible body region. The samples were stored in 95% ethanol and isotonic saline in equal proportions. This was believed to maintain the elastic properties of the bone for several weeks as shown by Ashman et al. in 1984 (55). The bone samples were tested with pulse transmission in 3 different directions namely, longitudinal, tangential and radial. The main finding of the experiment was that the elastic modulus in longitudinal direction of bone is higher when compared to the other 2 directions, which may imply greater resistance towards forces in that direction.

In 2003, Schwartz-Dabney and Dechow further investigated the variations of cortical bone properties throughout the dentate mandible (56). Similar methods and findings from an earlier study by Dechow et al in 1993 were reported. In addition, the authors have found that the direction of maximum stiffness, cortical thickness, cortical density, and elastic properties for the dentate human mandible demonstrate unique regional variation. For example, the stiffest cortical bone is found at sites in the ramus in the longitudinal direction, and that symphyseal cortical bone has the lowest density in the buccal region compared to other sites. This shows that the mandible bone is truly anisotropic, and non-homogenous.

Cancellous bone properties in the mandible have been investigated by Misch et al in 1999 and Giesen et al in 2001 (57, 58). Unlike the studies investigating properties of cortical bone, these authors used servo-hydraulic loads on mandible cancellous bone samples. The results differ significantly between the 2 studies. This might relate to the different sites of bone used in the study, Giesen et al. used bone in the condylar region, and Misch et al. studied bone in the mandibular body region (57, 58). However, the results not surprisingly show that cancellous bone has much lower elastic properties than cortical bone. Table 4.3 shows the values of cortical and cancellous bone reported in the literature.

Table 4.3 Bone Material Properties

Author	Bone	Elastic Modulus	Poisson's Ratio	Density
Dechow et al 1993 (54)	Cortical (Mandible Body)	11.3 – 19.4 GPa	0.33	1.768g/cm3
Schwartz-Dabney et al. 2003 (56)	Cortical (Mandible – whole)	12.7 – 22.8 GPa	0.3	1.747 – 1.968g/cm3
Misch et al. 1999 (57)	Cancellous (Mandible – body region)	96.23 MPa	NA	0.85 – 1.53g/cm3
Giesen et al. 2001 (58)	Cancellous (Mandible – condyle region	127 – 431 MPa	NA	2.127 – 2.146g/cm3

4.4.1.3 External Load / Post-operative bite force

Many studies have been carried out to determine the occlusal forces in healthy population as well those post-surgery. Studies by Ellis et al. in 1996 and Harada et al. in 2004 measures the occlusal forces in patients that underwent orthognathic surgery (59, 60). Their methods relied on static occlusal forces on pressure sensitive films and recorded the maximum forces obtained. The results show a range of 50 – 400N in different teeth locations with smaller forces recorded in the incisor region, and higher forces in the molar regions. This is related to the size of occlusal contact as well as the location of the teeth in the mouth. Molars are situated closer to the fulcrum & closing muscles of the jaw thus more force is generated in those areas.

In the study by Throckmorton et al in 1996 which showed post-operative bite forces at 6 months up to 2 years, the occlusal force distribution among teeth are approximately 50% in the molars, 35% in the premolars and 15% in the incisors (61). Similar results of force distribution are seen in the study by Ellis et al in 1996 (59).

There are no studies that have recorded immediate post-operative bite forces on patients undergoing orthognathic surgery as there are concerns of introducing complications such as bone fracture or hardware failure. Furthermore, patients might be in intermaxillary fixation (IMF) or on strict soft diet regime. Thus, it can only be estimated that at the early period of recovery, the bite forces would be lower than those reported in the literature.

Reproducing human bite forces in biomechanical studies must also be accompanied with a proper understanding of the oral physiology, and masticatory system. According to Table 4.2, the majority of the FEA studies applied vertical downward forces onto the occlusal surface. While this method might be reasonable in in-vitro studies, FEA studies should use the advantage of the computer software to simulate the physiologically functional human mandible. This is shown in the study by Stringhini et al. in 2016, which applied the forces in the direction of muscles of mastication in his model (47).

The human mandible movements are predominantly controlled by the muscles of mastication, which are the masseters, temporalis, lateral and medial pterygoid muscle. These muscles sizes are much bigger than the other depressor muscles as they are the main workforce of food chewing, and mastication. Greater forces are needed to elevate the mandible compared to depressing it.

During mastication, the maxillary teeth are static while the mandible is dynamic. The mandible bone undergoes a sagittal bending which is the result of forces of the muscle components and resulting forces on the condyle, and occlusal forces. This causes a tension force in the superior border and compressional force in the inferior border of the working side (62). This is true when looking at the mandible from a lateral perspective. However, the moments in the mandible is much more complex than that. This is because the muscles do not work in the same vector, or work identically with the contralateral side. The movement is also not purely vertical as there is some rotations in the condyles(63). Understanding the biomechanics of the mandible is crucial when designing biomechanical studies to simulate a normal physiologically functioning mandible.

The choice of fixation is relatively surgeon preference. Despite this, stability following orthognathic surgery is crucial. Fixation between bone segments must be able to withstand various internal and external forces to prevent excessive movements that can interrupt proper healing or even failure. Similar in mandible fractures, the fixation only aids in reducing the fracture, and holding the fractured segments together. They are not meant to withstand heavy forces within the period of bone healing, which takes approximately 6 weeks. A systematic review done by Joss et al. in 2009 for rigid fixations in mandibular advancements has shown that the short term relapse of the miniplates range is lower than the bi-cortical screws at 1.5 - 18% and 1.5 – 32.7% (64). Similar results are seen in long-term relapse where bi-cortical screws ranges between 2- 50% and miniplates ranges between 1.5 – 8.9%. The short term relapse is described by Schendel and Epker in 1980 as a cause of surgical technique while the long term relapse is related to the unbalanced stomatognathic system following surgery (65). Joss and Vasalli in 2009 then further concluded that the main factors for relapse are the amount of advancement, the choice of fixation, the mandibular plane angle, the stability of the proximal segment, soft tissues and muscles, and others (64).

Conclusion

Both in-vitro and finite element analysis (FEA) studies have shown comparable results in the biomechanical properties of fixation in mandibular osteotomies. With the use of FEA, it is possible to analyse not only stresses on the fixation, but also in the bone as well. This makes it a useful research tool. Generally, the use of bi-cortical screws in mandibular osteotomies have shown to provide the best stability among other fixation systems. Although miniplates are reported to perform less well than bicortical screws, they are still used widely due to the acceptable post – operative success. Having a rigid fixation may be advantageous in terms of stability, but how rigid must a fixation be, remains a question. Both rigid fixation and functionally stable fixation are used in mandibular osteotomies are able to provide good post-operative results but may still result in relapse in the long term. Thus, the question remains, is it necessary to have rigid fixation or is functionally stable fixation adequate for mandibular osteotomies?

5. MATERIALS AND METHODS

5.1 PART 1 – Model construction of the mandible

Computer software is used to form a 1:1 virtual model of the components used in this study. The components include a human mandible and the different internal fixations; 1.7mm miniplate with respective 5mm screws, 2.0mm miniplate with respective 5mm screws, and the 2.0mm bicortical lag screws. These softwares rely heavily on the specifications of the computer to perform well. For this study, the computer used, and specifications is listed as below:

Table 5.1 Computer Specifications

Model	Lenovo ThinkPad P50
Processor	Intel ® Core ™ i7-6700HQ CPU @ 2.60GHz
Operating System	Windows 10 Home Edition 64 bit, x64-based processor
Installed RAM	16.0 GB

This computer model is a certified solution partner for the software Solidworks, which is one of the software used for the finite element analysis simulation (66).

The following table lists the software used in this study.

Table 5.2 Lists of computer software used

Software	**Function**
InVesalius(67)	Reads CT Scan images (DICOM) and converts into 3-D polygon model. Allows user to choose areas of interest for conversion.
Netfabb(68)	Edits and repairs polygon surface model
Rhinoceros 5 (69)	Creates and manipulates NURBS surfaces. Allows inspection and editing of model.

FreeCad (70)	General purpose parametric 3D CAD modeler. Converts STL file into NURBS file.
Solidworks (66)	Creates, edit and manipulates 3D solid models.
Simulation (71)	An internal component of "Solidworks" that runs the Finite Element Analysis

5.1.1 Mandible Model

The model of the mandible was constructed using the CT Scan images in Digital Imaging and Communications in Medicine (DICOM) format. In order to form a solid 3-D model of the mandible, the DICOM images are initially converted into Non-Uniform Rational Basis Spline (NURBS) closed - surface polygons. It is then interpreted into a solid model of the mandible which is explained in detail below.

5.1.1.1 CT Scan of the Facial Bones

A pre-operative CT Scan of a 30-year-old, consenting female, who was undergoing surgery for excision of a maxillary tumour was used as the basis model of the mandible. The scan was exposed using a Lightspeed VCT Scanner (GE Medical Systems). The scan consisted of the full skull from the tip of cranium superiorly, including the orbits, zygomatic bone, maxilla and mandible, down to the hyoid bone inferiorly. It also included the cervical spine. The scan was formatted into DICOM images with 0.625mm splice thickness. This scan was chosen as the mandible had no previous history of trauma or pathologies, and has full dentition with minimal teeth restorations.

5.1.1.2 Stereolithography (STL) file creation (InVesalius)

CT Scan images were obtained from a CD provided by the Radiology Department of St. James's Hospital. The DICOM images were imported into the InVesalius Software. The default settings included all soft and hard tissues. The threshold value was changed to select only those areas of the CT scan that corresponded to Hounsfield Values representing bone. This resulted in highlighting the entire skull and cervical spine. To avoid complexity of editing, all the highlighted areas except for the mandible were deleted. This allowed the construction of a 3-D polygon model of the mandible and exported as a STL file. This model yields 171,042 triangles which provides sufficient anatomy accuracy.

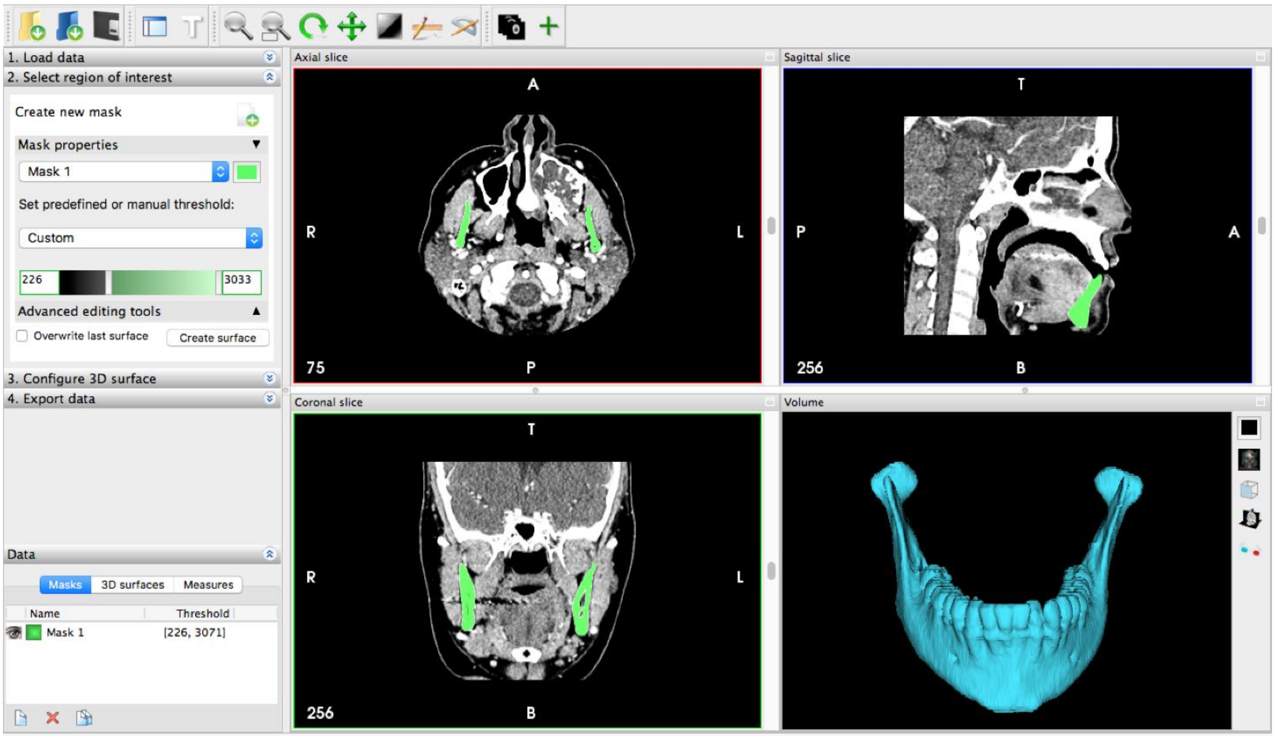


Figure 5.1 Forming the Mandible STL from patient's CT images in InVesalius

5.1.1.3 STL cleaning (Rhinoceros 5)

The STL model was opened in Rhinoceros (version 5.0.2) for inspection and editing. This model consisted of many ‘floating’ triangles inside and outside of the mandible model. These triangles were deleted in order to form a single surface polygon which is only compatible in Solidworks. Rhino allows the selection of the free triangles, and deletes them without interfering with the mandible model as shown in the figure below. Once the mandible model is cleaned, it is saved as an STL format again.

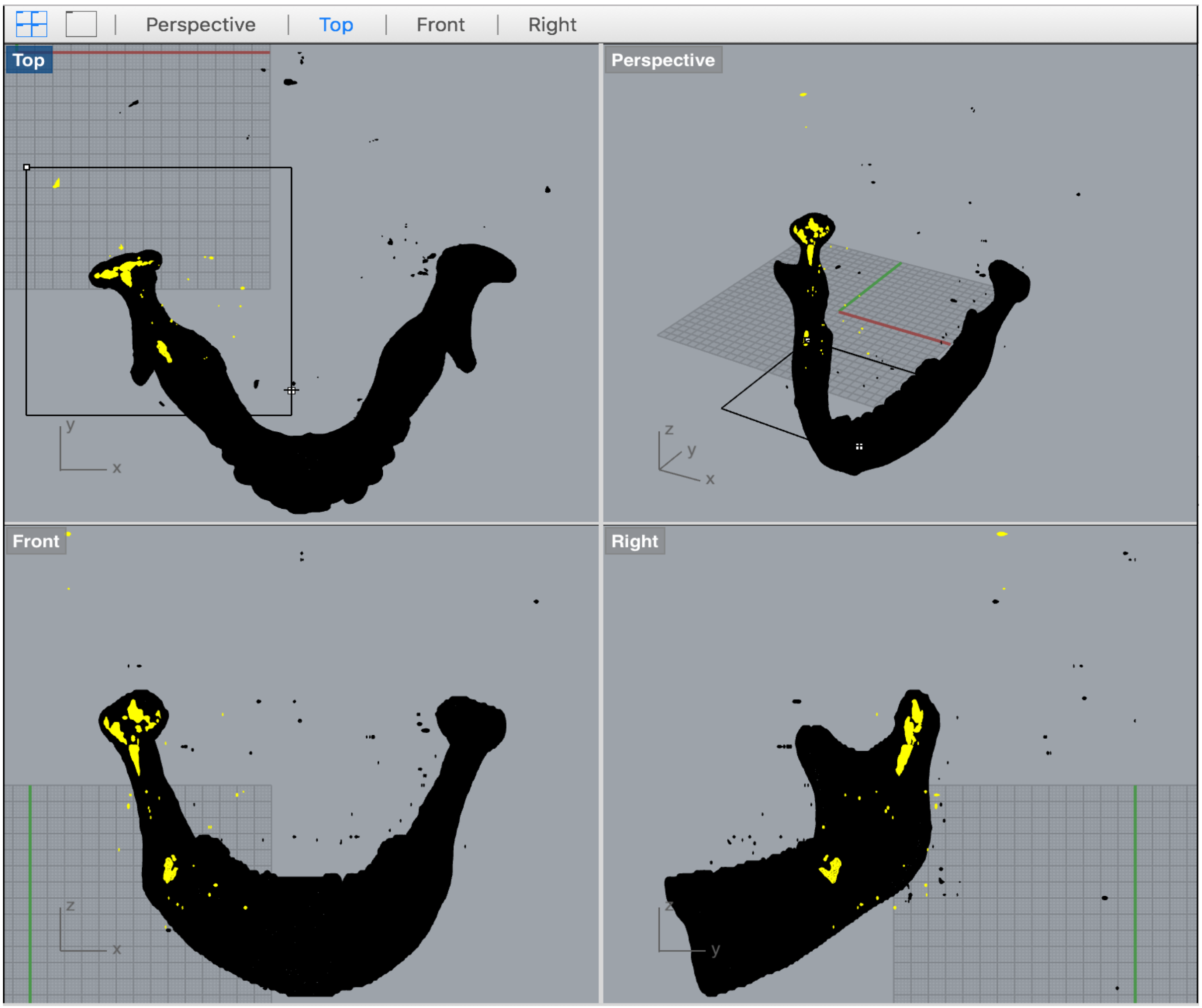


Figure 5.2 Removal of 'free' triangles using Rhinoceros

5.1.1.4 STL editing (NetFabb)

Further editing of the STL model was carried out in NetFabb. To create simplicity, and computer data management assuming symmetry of the model, it was split in half in the symphysis region and the left side discarded. Any holes or gaps present in the model were repaired automatically using the software. Figure 5.3 shows the result of editing. The repaired model was saved into STL format with resultant triangles of 15120. The repaired model was saved as a new STL file, made up of 15120 triangles. This was an anatomically accurate version of the right mandible. It then needs to be converted into a Non-Uniform Rational B-Splines (NURBS) surface model.

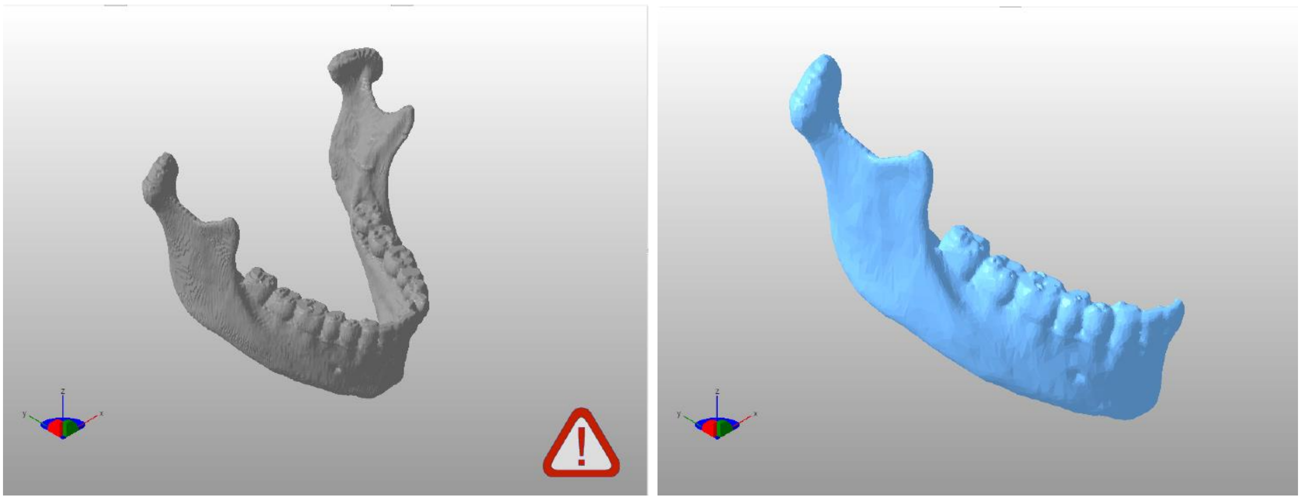


Figure 5.3 (Left) Before repair and split. (Right) Result of editing and split

5.1.1.5 NURBS creation (FreeCad)

The repaired STL model was imported into FreeCad software. This software allows conversion of the STL file into NURBS surface polygon format. This initially forms a NURBS model with multiple surfaces that are open and not attached to each other. However, SolidWorks prefer to work with solid models. Thus, this model needs to have the surfaces knitted together and this is done in the Rhinoceros software. Following formation of a single surface polygon, the model is saved as an Initial Graphics Exchange Specification (IGES) file format for use as a solid model.

5.1.1.6 Solid model (Solidworks)

The single surface NURBS model is imported into the SolidWorks software. SolidWorks interprets this closed-surface model as a solid. This model is now ready for assembly with other components. (Fig. 5.4)

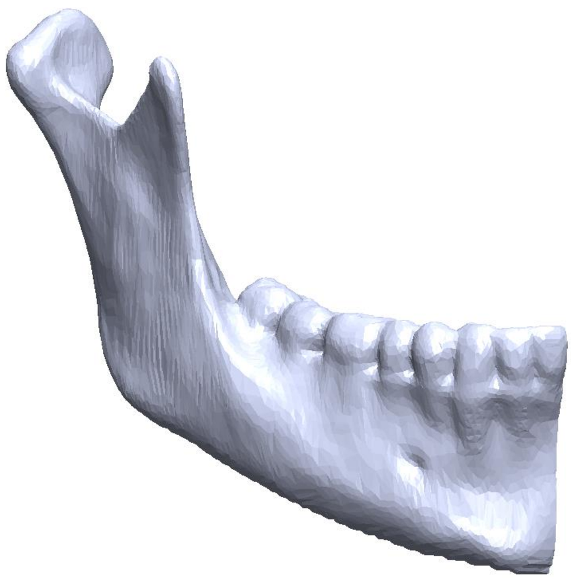


Figure 5.4 Solid Mandible Model

5.1.1.7 Cortical and Cancellous Bone Layers

The CT Scan splice thickness of 0.625mm was not able to accurately record the volume of cancellous bone in the mandible. This limitation caused formation of multiple small ‘floating’ triangles or cavities during the formation of the solid mandible model (Fig. 5.2). Due to this, they were removed during the cleaning process in Rhinoceros. However, the inclusion of a cancellous bone layer is crucial in this study and it was decided to create a cancellous bone layer, as explained in detail below.

The solid mandible model was edited to create a simplified cancellous bone part. The teeth were removed, and the file saved into a STL format (Fig. 5.5). This allows for manipulation in NetFabb to create an offset layer.

The STL model was offset to -2mm, meaning that it was shrunken 2mm inwards. The rough surface was smoothed using the smooth command which resulted in a cleaner model. Once the STL cancellous bone was satisfactory the same steps from NURBS creation to solid model was repeated. The final solid model is shown below (Fig. 5.6). It does not represent a true anatomical cancellous bone layer, but the model is satisfactory to provide adequate properties of a cancellous layer.

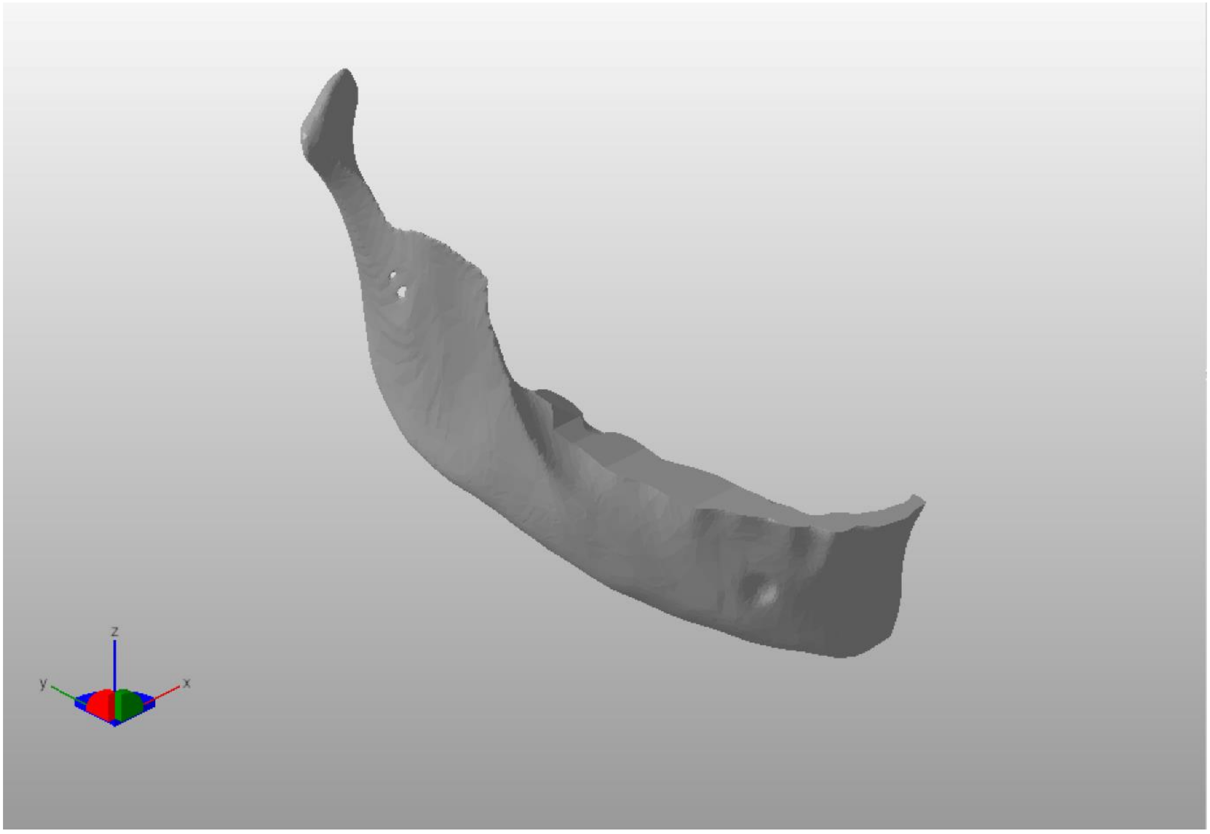


Figure 5.5 Solid mandible model following teeth removal

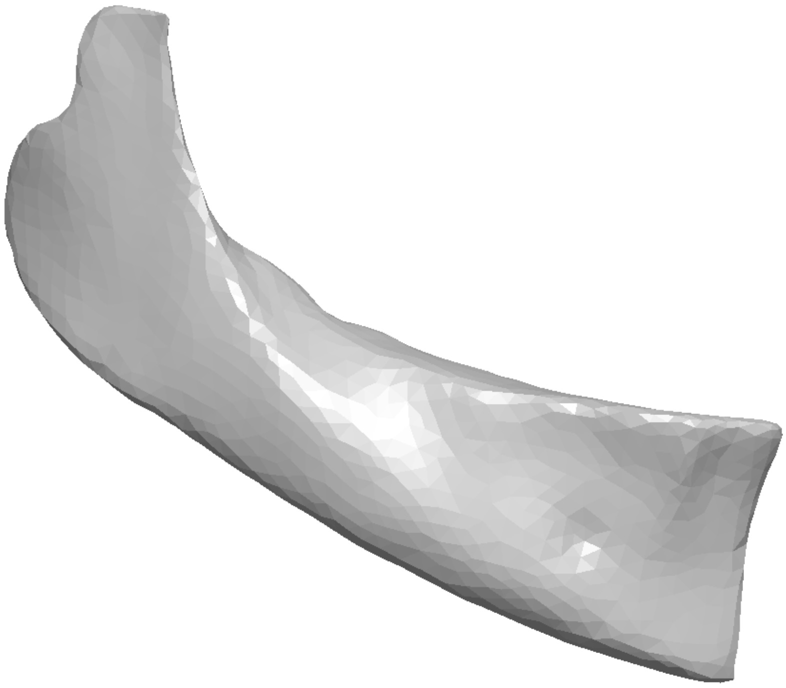


Figure 5.6 Solid Cancellous Bone

The initial mandible solid model and cancellous bone models were opened in SolidWorks for assembly. The mandible model was fixed in space to prevent any movements. The cancellous model was dragged into the main model and inspected in all 3 (X, Y, and Z) planes. This is to ensure that the entire cancellous model was submerged inside the main model uniformly. After positioning of the cancellous and main mandible model is satisfactory, the cancellous bone is fixed in space to prevent any accidental movements.

Currently, the cancellous model is interfering with the volume of the main mandible model. The main model is then edited to form a cavity according to the shape and volume of the cancellous model. After this is carried out, the assembly is saved as a SolidWorks Assembly format (.SLDASM). The cortical thickness is not uniform throughout the model. At the superior border on the buccal surface, it is ranged between 1.7mm – 2.0mm. At the inferior border, the cortical thickness ranges from 2.0mm – 2.5mm. This solid model consisting of a cortical and cancellous bone layer is now ready for further editing and manipulation (Fig. 5.7).

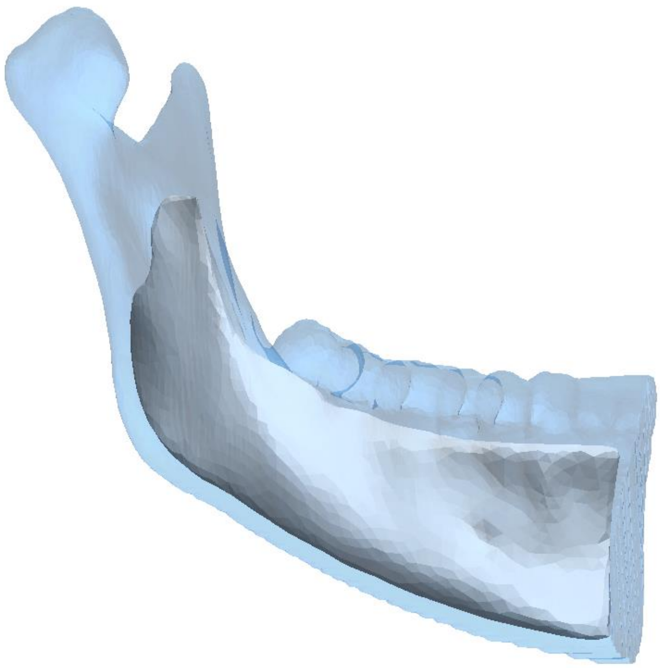


Figure 5.7 Assembly of main mandible with cancellous model

5.1.2 Sagittal Split Osteotomy

The chosen osteotomy design is the Sagittal Split Osteotomy (Obwegeser, DalPont, Hunsuck, and Epker Modification) as it is a versatile technique that allows advancement and setback of the proximal and distal segments. It is also one of the most common techniques used in mandibular orthognathic surgery.

The current mandible model is opened in SolidWorks to carry out the osteotomy cuts. It is viewed from the Top Plane. The osteotomy cuts are drawn onto the plane and extruded through the mandible model. The lingual cut is made superior to the lingula and the buccal cut is placed mid buccal of the second molar. This results in the formation of the “Distal Segment” consisting of both cortical and cancellous bone solid parts. As these parts are referenced together, they are saved separately as solid parts. The same is carried out for the proximal mandible segment.

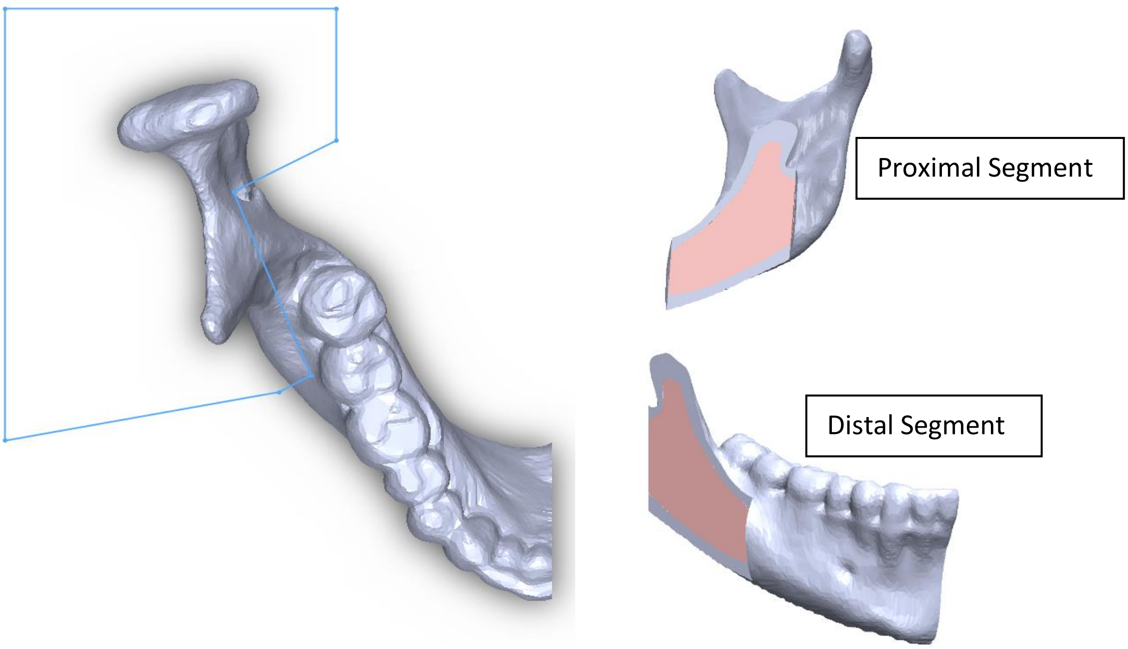


Figure 5.8 (Left) Sketch of Sagittal Split Osteotomy cut (Obwegeser, DalPont, Hunsuck, and Epker Modification). (Upper Right) Proximal Segment – Lingual view, (Lower Right) Distal Segment – Buccal view. Grey – Cortical bone, Pink – Cancellous bone.

5.1.3 Internal Fixation

The 3 types of internal fixation were created in SolidWorks. Only conventional miniplate designs were used in this study. The dimensions were taken from the manufacturer Stryker®.

5.1.3.1 1.7mm Miniplate and screws

1.7mm refers to the diameter of the screw threads. This type of fixation is commonly used in the midface region. It does not have any gap between holes but can range from 4 to 8 holes. A circle of 3mm diameter was drawn and protruded 0.7mm to form the thickness of the plate. A second circle was drawn in the same centre of 1.7mm diameter to form the screw hole. This circle was then protruded 5.7mm to form a 5mm length screw. The connector length between holes is 1mm. For use in longer mandibular advancements, a 6 or 8-hole plate is used. As the screws are rigidly attached to the bone during FEA process, the screw threads were omitted to reduce complexity of the model. The drawing schematics of the miniplate and screws is shown in Fig 5.9.

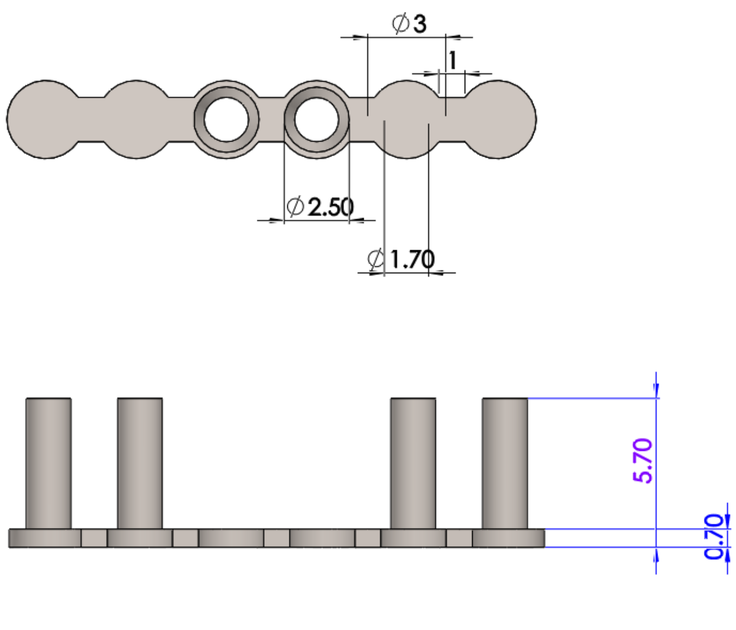


Figure 5.9 Schematic Diagram of 1.7mm miniplate and screws

5.1.3.2 2.0mm Miniplate and Screws

The 2.0mm miniplate and screws are developed in the same manner as the 1.7mm miniplate. However, the 2.0mm miniplate comes in variable gap length (between the 2nd and 3rd screw hole) from no gap to 4.4mm or an 8mm gap. The 2.0mm refers to the diameter of screw threads and developed as cylinders. The outer circle diameter is 4.5mm and has a thickness of 1mm. The connector length between screw holes is 2mm. (Fig. 5.10)

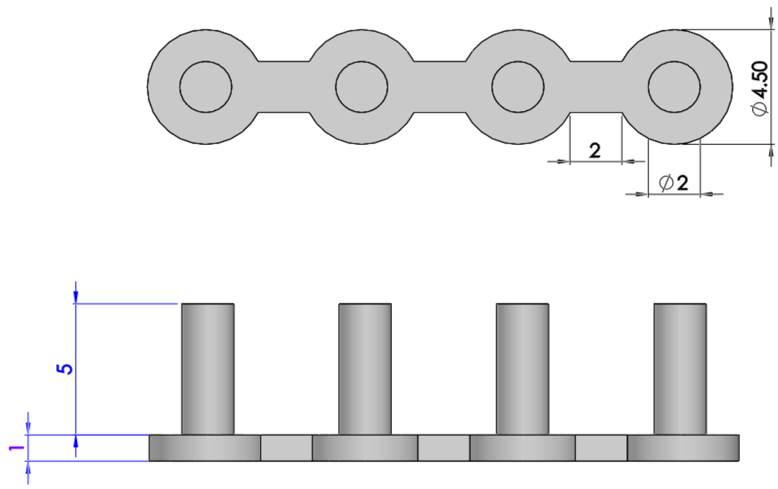


Figure 5.10 Schematic Diagram of 2.0mm miniplate and screws

5.1.3.3 2.0mm Bicortical Screws

Three bicortical screws were developed in Solidworks. The screw threads are of 2.0mm in diameter. The screw head has a diameter of 3mm. Initially, two screws had 15mm length while one screw was 11mm in length. (Fig. 5.11) The lengths of the screws will change when assembled with the mandible model to ensure both cortices of the bone is penetrated.

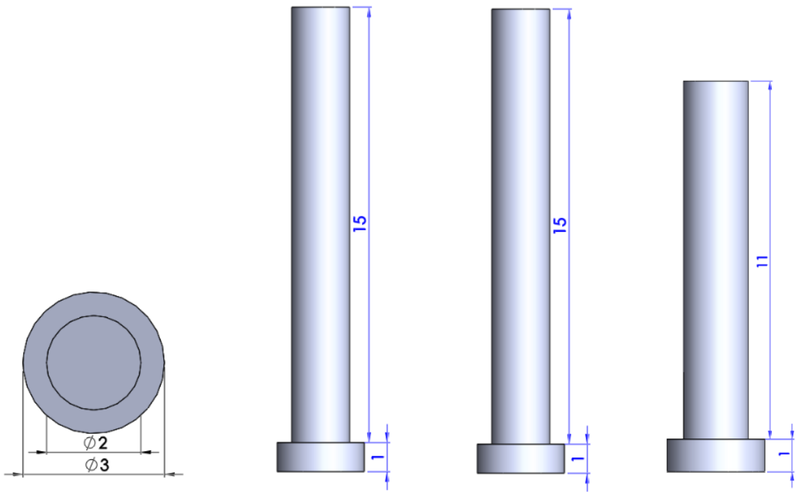


Figure 5.11 Schematic diagram of 2.0mm bicortical screws

5.1.4 Model Assembly

A total of 9 models were constructed from the 3 fixations techniques and 3 osteotomy movements.

Table 5.3 Model Assembly

FIXATION GROUP	1.7mm miniplate	2.0mm miniplate	2.0mm Bi-cortical Screws
3mm Setback	Model 1	Model 2	Model 3
3mm Advancement	Model 4	Model 5	Model 6
7mm Advancement	Model 7	Model 8	Model 9

The components included in the assembly are:

- Proximal Cortical and Cancellous Bone
- Distal Cortical and Cancellous bone
- Fixation type

The bone components were opened in SolidWorks. For the proximal part, the cortical and cancellous bones were brought together, and 3 matching faces mated coincidentally to fix the components together. This process is repeated with the distal cortical and cancellous bones. The result are 2 bone segments assembled; the proximal and distal segments.

Next, the lingual surface of the proximal cancellous bone is then mated with the buccal surface of the distal cancellous bone. This allows the two bone segments to 'stick' together. The anterior region of the proximal bone is mated with the corresponding surface of the distal segment but distanced apart according to the desired movement. For the setback movement, 3mm of the buccal and lingual bone in the proximal segment was cut out to allow proper approximation of the distal segment.

Next, the fixation is applied on to the mandible to 'fix' the two segments together. The plate is carefully placed, as close as possible to the buccal bone. The screw is embedded inside the bone. At this moment, the screws are interfering with the mandible. A cavity command was carried out to correspond to the screw shape and length to avoid interference in the model.

The 9 assembled models are shown below.

5.1.4.1 Model 1

3mm of the buccal and lingual part of proximal bone was removed and the distal bone setback 3mm. The 1.7mm miniplate has 5 holes with the 3rd hole acting as a connector. Two 1.7mm screws of 5mm in length were placed on each bone segments. The plate was slightly bent to allow the best adaptation to the bone surface without intruding into the bone and causing interferences. (Fig.5.12)

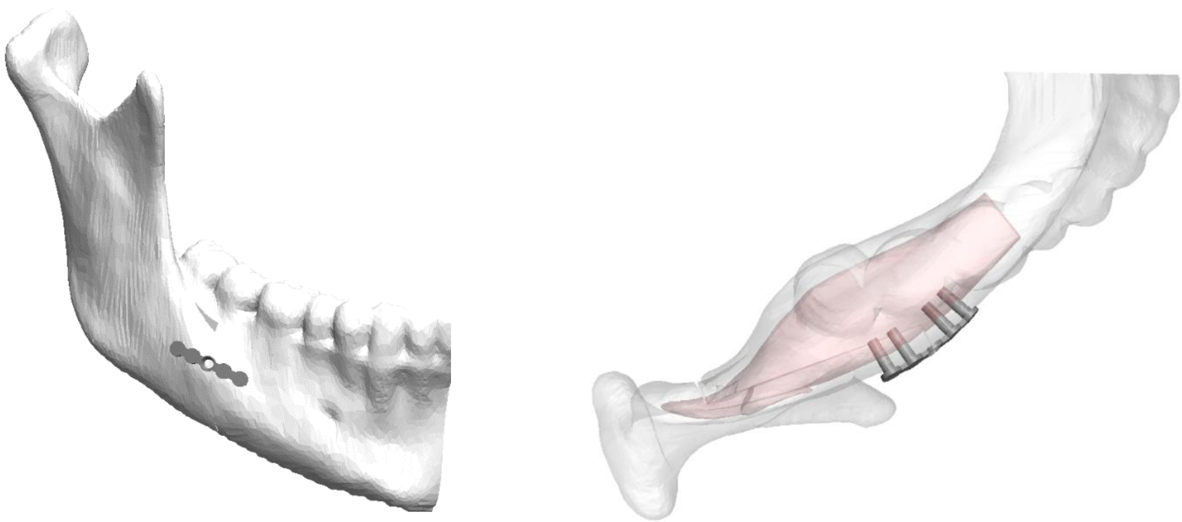


Figure 5.12 Miniplate 1.7mm at 3mm Setback. (Left) Buccal View, (Right) Transparent Superior View showing penetration of 5mm mono-cortical screws

5.1.4.2 Model 2

This is a duplication of Model 1. The 2.0mm miniplate without connector gap replaces the 1.7mm miniplate. Two 2.0mm screws with 5mm length were placed on each bone segments. The plate was slightly bent to allow best adaptation to the buccal bone surface without causing any interferences in the model. (Fig. 5.13)

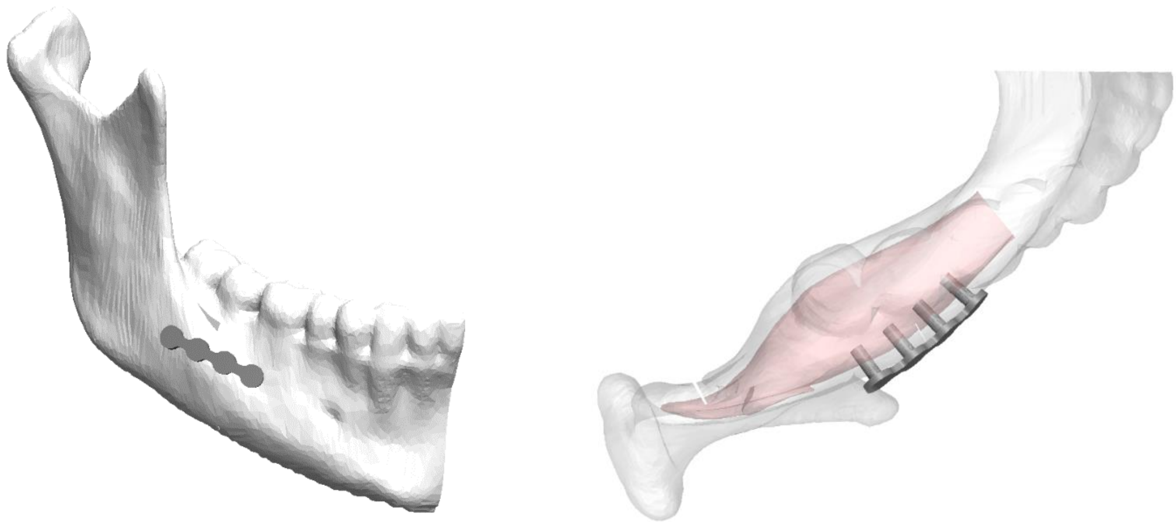


Figure 5.13 Miniplate 2.0 mm at 3mm Setback. (Left) Buccal View, (Right) Transparent Superior View showing penetration of 5mm mono-cortical screws

5.1.4.3 Model 3

This is a duplication of Model 1. Three bi-cortical screws replace the 1.7mm miniplate. The screws are labelled as 'superior-proximal', 'superior distal', and 'inferior distal'. The screws are arranged in an 'Inverted-L' position with two screws superiorly at 15mm in length and 1 screw inferiorly at 11mm in length. These screws penetrate the lingual cortex of the distal bone. The location of the screws is intended to be away from the roots of the posterior teeth and the inferior dental nerve. (Fig. 5.14)

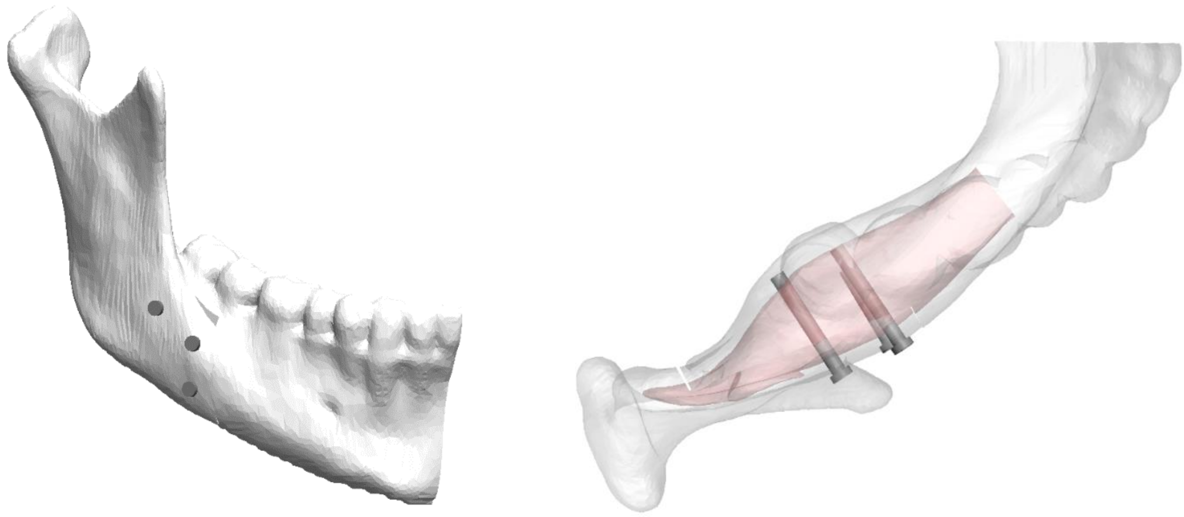


Figure 5.14 Inverted-L pattern of 2.0mm Bi-cortical Screws at 3mm Setback. (Left) Buccal View, (Right) Transparent Superior View showing penetration of the Bi-cortical screws

5.1.4.4 Model 4

The distal segment was advanced 3mm forwards from the proximal segment. A 6 hole 1.7mm miniplate was used with the 3rd and 4th hole acts as the connector. 2 screws of 5mm length was applied on each bone segment. The plate was slightly bent to allow best adaptation to the buccal bone surface. (Fig. 5.15)

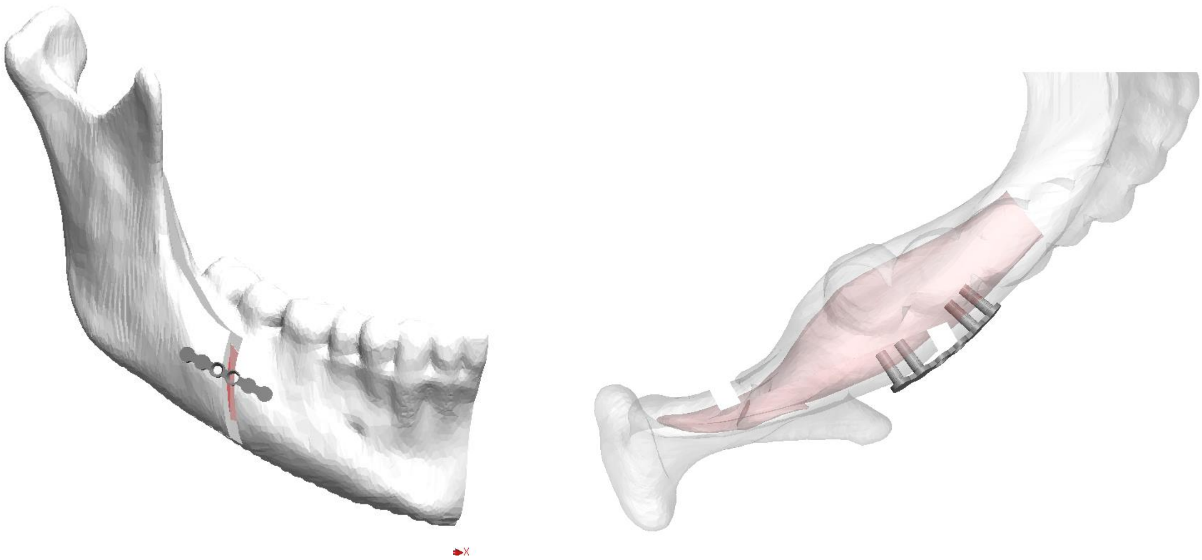


Figure 5.15 Miniplate 1.7mm at 3mm Advancement. (Left) Buccal View, (Right) Transparent Superior View showing penetration of 5mm mono-cortical screws

5.1.4.5 Model 5

This is a duplication of Model 4. The 1.7mm miniplate is replaced with the 2.0mm miniplate with 4.4mm connector gap between the 2^{nd} and 3^{rd} hole. 2 screws of 5mm length is placed on each side and the plate was slightly bent to allow best adaptation to the bone surface. (Fig. 5.16)

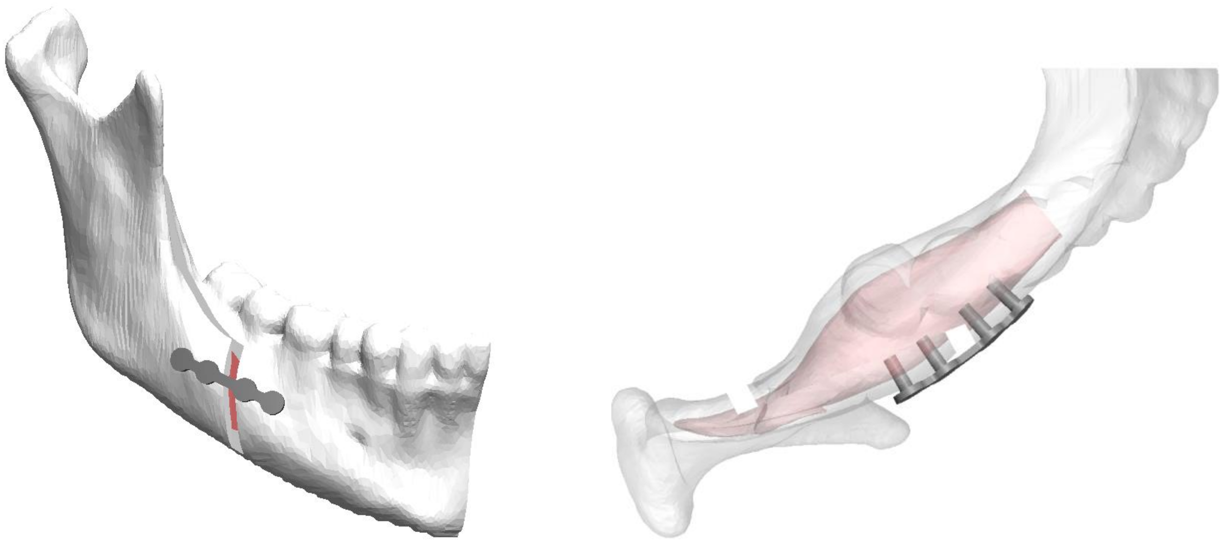


Figure 5.16 Miniplate 2.0 mm at 3mm Advancement. (Left) Buccal View, (Right) Transparent Superior View showing penetration of 5mm mono-cortical screws.

5.1.4.6 Model 6

This is also a duplication of model 4. The 1.7mm miniplate is replaced with 3 bi-cortical screws in an inverted-L pattern. The screws are 16mm, 17mm and 11mm in length for the superior-proximal, superior distal and inferior distal screws respectively. (Fig. 5.17)

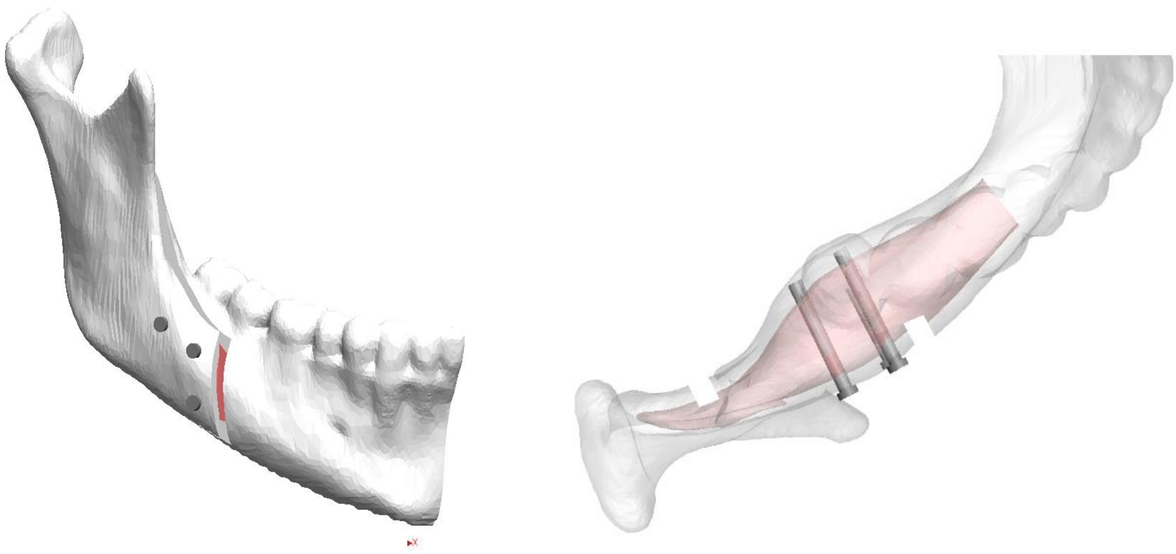


Figure 5.17 Inverted-L pattern of 2.0mm Bicortical Screws at 3mm Advancement. (Left) Buccal View, (Right) Transparent Superior View showing penetration of the Bicortical screws

5.1.4.7 Model 7

The distal segment was advanced 7mm forwards from the proximal segment. A 7 hole 1.7mm miniplate was used with the 3rd, 4th, and 5th hole acts as the connector. 2 screws of 5mm length was applied on each bone segment. The plate was slightly bent to allow best adaptation to the buccal bone surface. (Fig. 5.18)

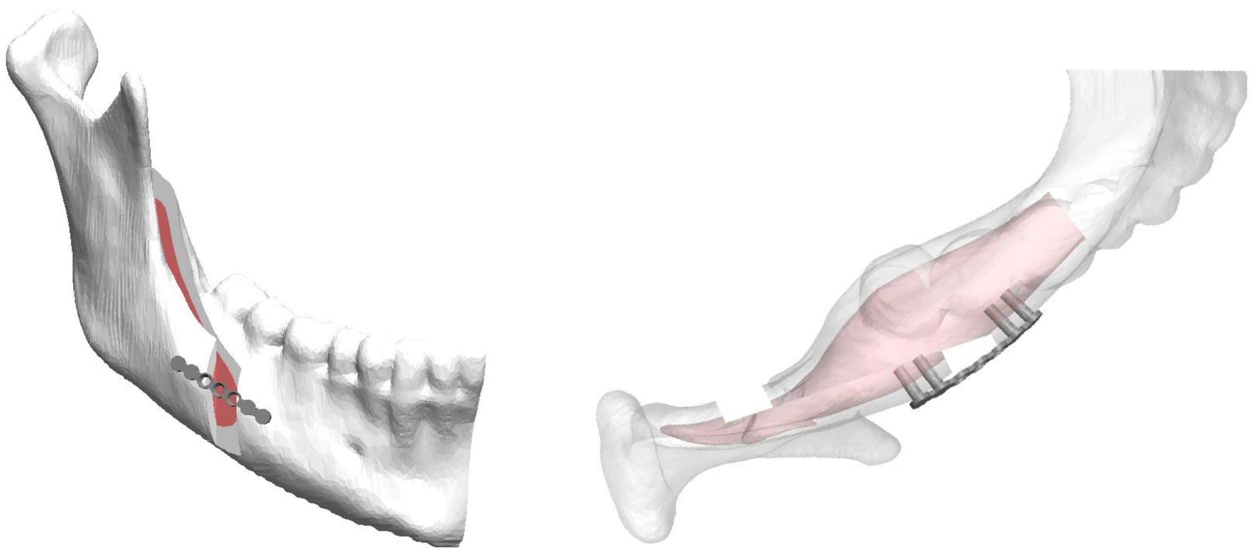


Figure 5.18 Miniplate 1.7mm at 7mm Advancement. (Left) Buccal View, (Right) Transparent Superior View showing penetration of 5mm mono-cortical screws

5.1.4.8 Model 8

This is a duplication of Model 7. The 1.7mm miniplate is replaced with the 2.0mm miniplate with 8.2mm connector gap between the 2^{nd} and 3^{rd} hole. 2 screws of 5mm length is placed on each side and the plate was slightly bent to allow best adaptation to the bone surface. (Fig. 5.19)

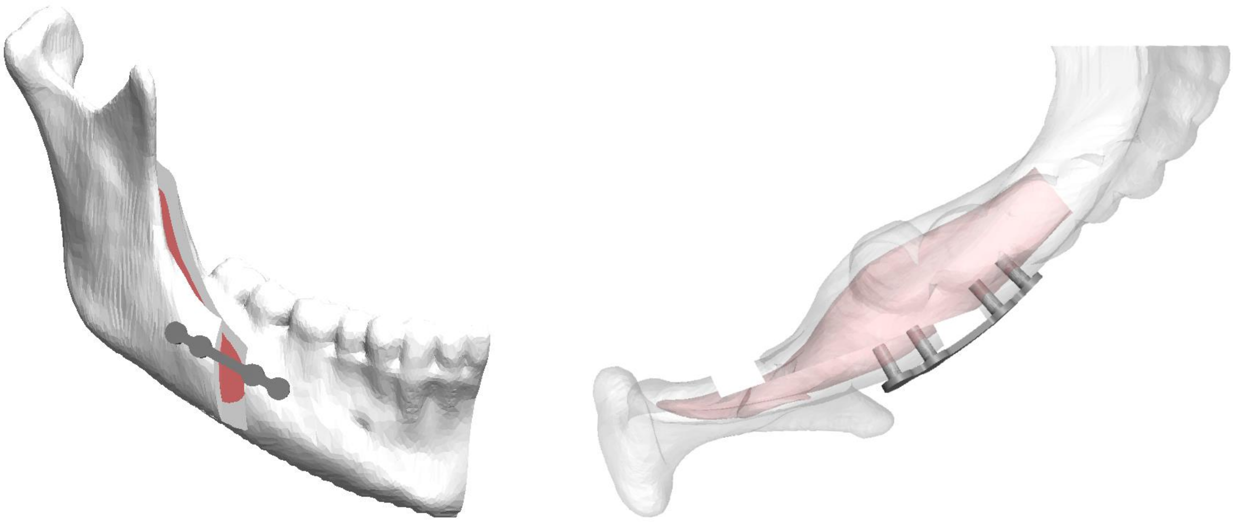


Figure 5.19 Miniplate 2.0 mm at 7mm Advancement. (Left) Buccal View, (Right) Transparent Superior View showing penetration of 5mm monocortical screws.

5.1.4.9 Model 9

This is also a duplication of model 7. The 1.7mm miniplate is replaced with 3 bicortical screws in an inverted-L pattern. The screws are 13mm, 15mm and 10mm in length for the superior-proximal, superior distal and inferior distal screws respectively. (Fig 5.20)

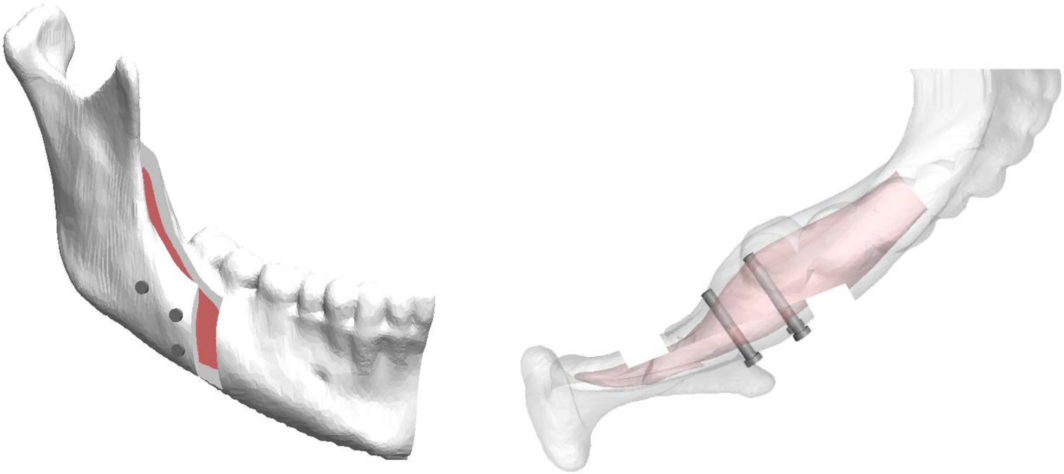


Figure 5.20 Inverted-L pattern of 2.0mm Bi-cortical Screws at 7mm Advancement. (Left) Buccal View, (Right) Transparent Superior View showing penetration of the Bi-cortical screws

5.2 PART 2 – FINITE ELEMENT ANALYSIS SIMULATION

Prior to running the FEA analysis, the mandible model assembly needs to have various properties defined. These properties include study type, material properties, contacts and connections, fixtures, loads and mesh. Each properties will be discussed in detail.

5.2.1 Study type

The study type chosen is non-linear. This is because organic materials like bone are non-linear in nature and the non-linear simulations usually produce more accurate results with bending, when compared to linear static studies.

5.2.2 Material properties

Simulation in Solidworks requires the user to define the material properties for components used in the study. The software provides a vast materials library, but does not contain any human or organic materials like bone or other tissues. The material properties for cortical and cancellous bones were average values derived from previous studies(56, 58-61). Table 5.4 defines the material properties used in the study.

Table 5.4 Material Properties

Material	**Elastic Modulus**	**Poisson's Ratio**	**Density**
All Fixation devices (Ti-6AI-4V)	104.8 GPa	0.31	4428.8 kg/m^3
Cortical Bone	17 GPa	0.3	2000 kg/m^3
Cancellous Bone	300 MPa	0.3	1500 kg/m^3

5.2.3 Contact properties

The surfaces between components are defined as no-penetration, allow penetration, and bonded. The contacts between the proximal and distal part of mandible are defined as no penetration, as this study assumes the parts are not healed or fused together (immediately post-operative), and also allows them to move or slide amongst each other. The contact surfaces between cortical and cancellous bones of each proximal and distal parts are defined as bonded, as this occurs naturally. The contact surfaces between the fixation device and the rest of the components are defined as bonded, as it is assumed that they are rigidly fixed together, since the screw threads are omitted in the design. The contact properties remain constant in all models and simulations.

5.2.4 Boundaries and Restraints

Boundaries and restraints in the model are applied to regions where movement is restricted. The condyle is fully restrained in all dimensions. This is anatomically and physiologically not. Due to the nature of the study, where only the mandible is present, the condyle was required to be fixed, to prevent the whole model from ‘floating’. The symphysis surface is restrained with a slider/roller fixture, which permits movement in the vertical plane but not in the horizontal plane. This allows simplification of the model, where a half mandible can be used and symmetry is assumed. The computer can then more easily manage the data load. The incisor and molar occlusal surface are also fixed depending on the type of simulation used in the study. When not in use they can be suppressed. The fixtures are shown as green arrows on the model.

5.2.5 External load

Load is applied onto the model to simulate bite forces during function. The forces applied for incisal region are 50, 75 and 100N, whereas forces for molars are 100, 200 and 300N. Specific loading conditions is explained in the next section (5.3.1).

5.2.6 Discretization (mesh creation)

All assembled models undergo meshing in Simulation SolidWorks. The final meshing settings and values were obtained once mesh convergence testing was carried out as explained in 4.3.2.

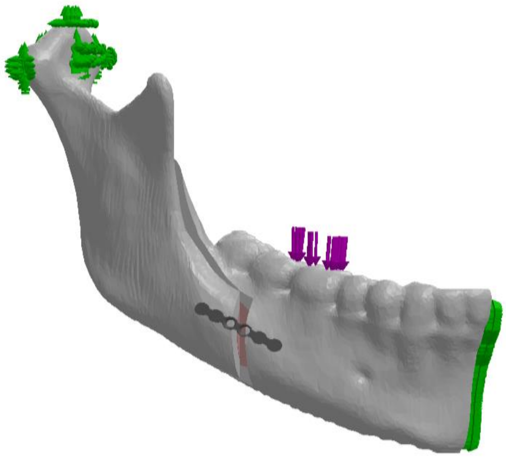
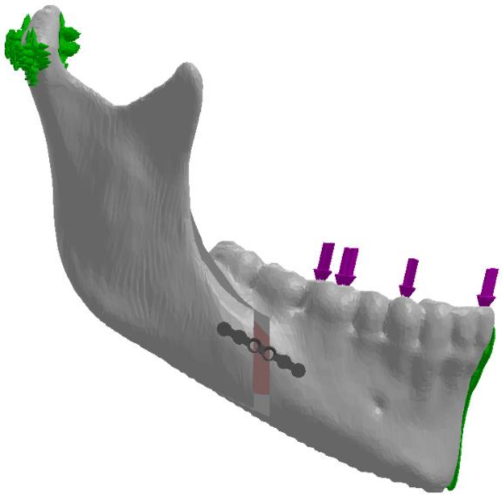
5.3 PART 3

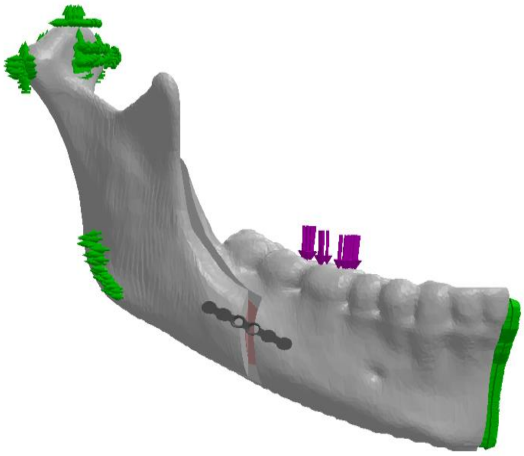
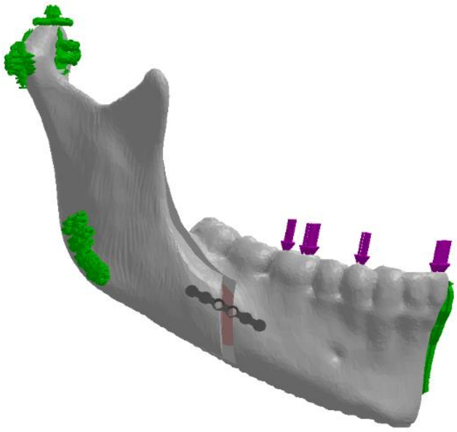
5.3.1 Simulation Models

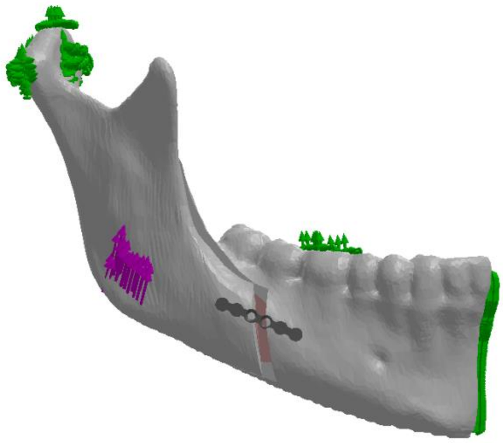
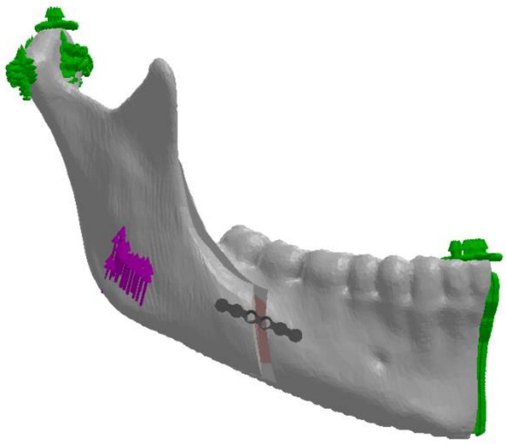
The aim is to compare various simulation models as described in previous FEA studies. A total of 18 Simulation models were constructed using Models 4, 5, and 6. A total force of 100N was applied in each model. Stresses in fixations, and maximum displacement of bone segment was recorded and compared.

The figures in the table below show the various simulation models on Model 4 (Fig 5.15, page 47). Each simulation model was duplicated for the 2.0mm miniplate (Model 5) and 2.0mm Bicortical Screws (Model 6).

Table 5.5 Simulation model description and illustration - (Green Arrows = Fixture/Restraint, Purple Arrows = Load/Force)

Model	Features	Figure
A	This simulation model represents the model commonly used in in-vitro studies and some FEA studies. The condyle is fixed in all directions and a vertical force of 100N is applied onto the occlusal surface of the 1st molar tooth.	
B	This simulation model is similar to the conventional model A. However, 2 additional force loads are applied onto the premolar and incisor region. A force of 100N is applied in the ratio of 50:35:15 in the molar, premolar and incisor region respectively. This aims to simulate the spread of bite forces onto the whole occlusal surface as described in bite forces studies(59, 61)	

C	This is a duplication of model A. An additional restraint is applied to the angular region to simulate the pterygo-masseteric sling. This type of restraint is also applied in most in-vitro studies, as it stabilizes the proximal bone segment and prevents clockwise rotation during loading. A single vertical force is applied on the occlusal surface of the 1st molar.	
D	This model is a duplication of model C with application of forces from model B.	

E	This model implies forces generated by the masseter muscle. The condyle is fully restrained as in other models. The symphysis region has a roller / slider restraint and the occlusal surface of the molar has a fixed restraint to simulate posterior teeth biting. The force is generated in the angle region of the mandible perpendicular to the occlusal line.	
F	This is a duplication of Model E. The only difference is the restraint on the molar is suppressed and a restraint on the incsior tooth is placed. This simulates an incisal biting condition.	

5.3.2 Mesh Convergence Test

The aim for the convergence test is to determine the best mesh settings and number of elements that can produce the most accurate results. A coarse mesh with less elements would not yield as true a predicted behaviour of a material, as a finer mesh with more elements would. Convergence is achieved when the analysis reaches a solution independent of the mesh size. In reality, the solution will always increase with more number of elements. Thus, a limit of 5% difference in successive results dictates acceptable result and mesh size.

Refining the mesh by reducing element sizes of the whole model is one way of producing more accurate results. However, this method is time consuming, uses more computer resources, and areas further from the point of interest does not require finer mesh. Alternatively, local refinement in Simulation allows the control of element sizes in areas of interest. In this study, a series of analysis was carried out on Model 4 (Fig 5.15, page 47) at 100N force, as this model consisted of the smallest fixation technique which is the 1.7mm miniplate. A curvature based mesh was chosen for best adaptation to the mandible geometry. The first analysis (Mesh A) consisted of default maximum element size of 6mm and minimum element size of 1mm. This is to determine the area of high stress, which is anticipated to be in the miniplate. Further analyses applied local refinement to the miniplate without changing the overall element size. Local refinement of element size in the miniplate is reduced until 5% difference in successive results, which indicates convergence.

Table 5.6 Mesh model element sizes and local refinement

MESH	Max element size	Min Element Size	Local Refinement in 1.7mm miniplate
A	6	1	-
B	6	1	0.5
C	6	1	0.35
D	6	1	0.3

5.3.3 Comparison of fixation techniques in 3 different Mandibular Movements

Results from 5.3.1 and 5.3.2 was used to determine the best simulation model and mesh setting for further analysis and comparison of the 3 fixation techniques in various mandibular sagittal split movements. All three fixation techniques (1.7mm miniplate, 2.0mm miniplate and 2.0mm Bicortical Screws) were compared at 3mm Setback, 3mm Advancement and 7mm Advancement movements. The respective model assemblies, which are shown in section 5.1.4 underwent FEA simulation at forces 50, 75 and 100N in the incisal region and 100, 200 and 300N in the molar region.

The amount of maximum displacement and its location is recorded in all simulations. The amount of maximum Von Mises stress and Equivalent Strain in each fixation techniques, and surrounding cortical and cancellous bone are recorded. Statistical analysis was carried out to compare the maximum stresses and displacements between the 1.7mm miniplate with the 2.0mm miniplate and 2.0mm bi-cortical screws. Normality of data was tested using the Shapiro-Wilk's test. Based on the normality of the data, Student's T- test was carried out if data was normally distributed, and Wilcoxon rank sum test when data is not normally distributed.

6. RESULTS

6.1 Comparison of simulation models

The maximum displacement of the mandible model and von Mises stresses in all three fixation techniques for each simulation models are shown in the tables 6.1 and 6.2.

The highest amount of displacement and stress is seen in the simulation Model B where the bite forces are spread onto the incisor, premolar and molar teeth. Maximum displacements for Models A to D are seen in the lower border of symphysis of the distal bone segment, while Models E and F have their maximum displacement in the angle region of the proximal bone segment (Table 6.1). This indicates that the location of displacement depends on the area of force application.

The least displacement for the 2.0mm miniplate and bi-cortical screws fixation is seen in Model E, and Model F for 1.7mm miniplate fixation. Model F also records the least stress values for all fixation techniques.

Table 6.1 Maximum Displacement (mm) in Simulation Models

	Fixation Type		
Simulation Model	1.7mm Miniplate	2.0mm Miniplate	2.0mm Bi-cortical Screws
A	2.621	2.416	1.805
B	4.184	3.432	2.37
C	1.179	1.05	0.6089
D	2.099	1.641	0.8731
E	0.0624	0.1186	0.09
F	0.04695	0.1258	0.1045

Table 6.2 Maximum Von Mises Stress (MPa) in Fixations

	Fixation Type		
Simulation Model	1.7mm Miniplate	2.0mm Miniplate	2.0mm Bi-cortical Screws
A	864.2	593.6	212.2
B	1062	842	296.1
C	829.3	560	206.2
D	1040	786.9	288.6
E	265.6	132.4	63.84
F	218.2	123.3	57.18

6.2 Mesh Convergence Test

A series of Finite Element Analysis were carried out on Assembly Model 4 (Fig. 5.15, page 47) at 100N force, to determine the best mesh settings. A total of 4 simulations were carried out to show convergence of results with less than 5% difference compared to the previous coarser mesh. Mesh A have shown that the high stresses in the model is concentrated in the miniplate. Mesh B, C, and D then underwent local refinement, where only the element size in the miniplate alone was reduced and other parts of the model remained constant. This is to ensure that the stress results in the miniplate become more accurate. The total number of elements generated, and maximum Von Mises Stress in the 1.7mm miniplate fixation was recorded as shown in Table 6.3.

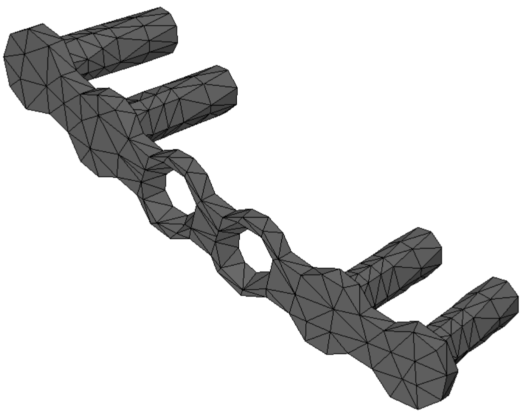
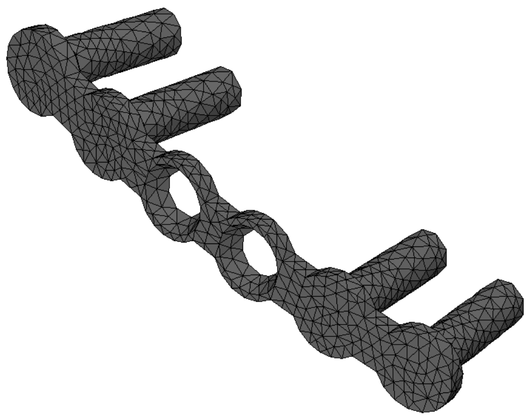
A table of figures showing local refinement in the miniplate is given in Table 6.4. Increased refinement in mesh size shows improvement in geometry accuracy as seen in Models A to D. Model A had no local refinement while models B, C, and D had local refinement of 0.5mm, 0.35mm and 0.3mm respectively. The models with local refinement exhibits better geometry accuracy, where the curves on the miniplate, and screws appear to be smoother.

Mesh models' C and D recorded a 4% difference in the stress result. The results in these models have reached convergence and Model C has the maximum element size to obtain as close accurate result required in the analysis. Thus, Model C's mesh setting was chosen to be used in further simulations in this study, as it reduces demand of computer resource and increases efficiency in running the analysis.

Table 6.3 Effect of mesh model refinement on stress and displacement values

Mesh	Max. Element Size (mm)	Min. Element Size (mm)	Local Refinement (1.7mm Miniplate)	Elements	Stress (MPa)	% difference with previous mesh
A	6	1	N	119480	265.6	-
B	6	1	0.5	125560	286.4	7.8%
C	6	1	0.35	136450	307.5	7.3%
D	6	1	0.3	143608	320.3	4%

Table 6.4 Figures of the 1.7mm miniplate with local refinement

 Mesh A	 Mesh B
 Mesh C	 Mesh D

6.3 Comparison of fixation techniques in various mandibular movements

Based on results in 6.1 and 6.2, simulation Models E and F was chosen to be used for further analysis and comparisons of the 3 internal fixation techniques (6.3.1 – 6.3.6). Model E was used to simulate molar biting and model F simulates incisal biting.

6.3.1 Setback 3mm – Stress distribution in models and fixation techniques

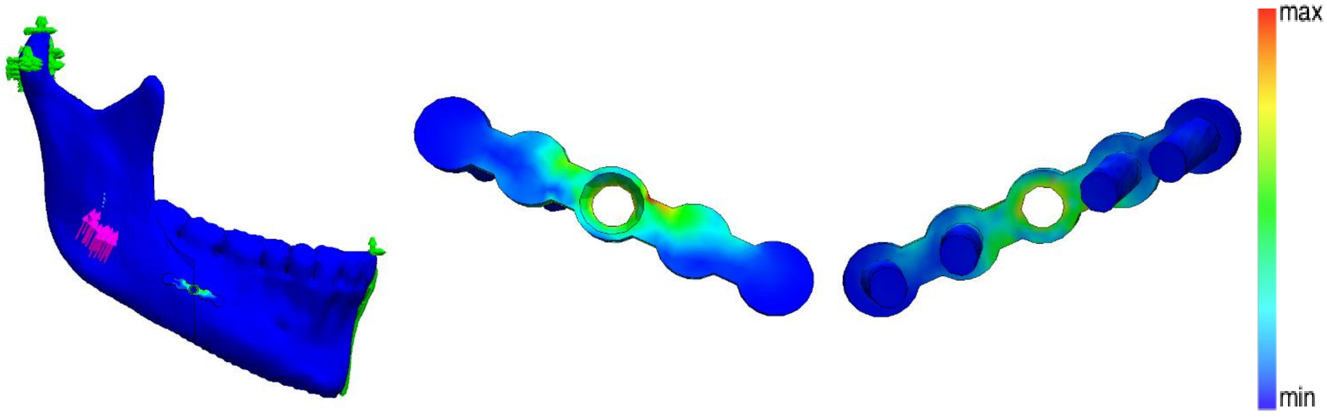
The distribution of stresses in the models are shown in Figures 6.2 – 6.19. Maximum stresses are represented in red, and minimum stresses are represented in blue. As these values vary between models, the scales used to assign colours are also different between models.

In Model 1 (figures 6.1 – 6.7) with the 1.7mm miniplate fixation, the stress is generally distributed in the miniplate connector region at the 3rd hole. Maximum stress is seen in the superior – distal region of the 3rd mini-plate hole. Minimum stresses are seen in the screw threads. This pattern is relatively constant with either incisal or molar restraint. The maximum stress regions increase with increasing force magnitude, with higher stresses developing in the inner circle region of the 3rd hole at 300N force.

The 2.0mm miniplate fixation in Model 2 (figures 6.8 – 6.13) shows similar stress distributions as Model 1. The stress is distributed along the connector region of the miniplate and maximum stress is concentrated at the superior - distal part of the connector and mesial region of the 3rd hole. In addition, models with force magnitude of 50 to 100N with incisal restraint, high stresses are also seen in the inferior – proximal part of the miniplate connector near the 2nd hole. However, this is not seen in the molar restrained models.

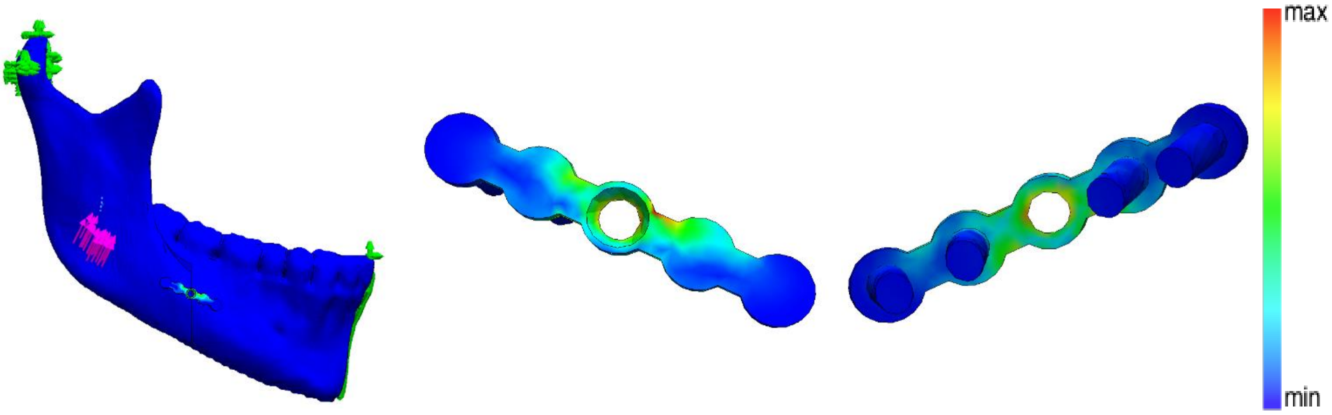
Model 3 fixation technique involves 3 Bi-cortical screws arranged in an inverted-L orientation (figures 6.14 – 6.19). The screws are labelled as Superior – proximal (1), Superior

– distal (2), and Inferior – distal (3). Stresses are distributed in the screw threads of all screws and are generally concentrated at the cortical-cancellous bone interface. Screw 2 and 3 shows more stress distribution with maximum stress in the lingual region of the inferior distal screw. This stress distribution pattern is relatively constant throughout the model.



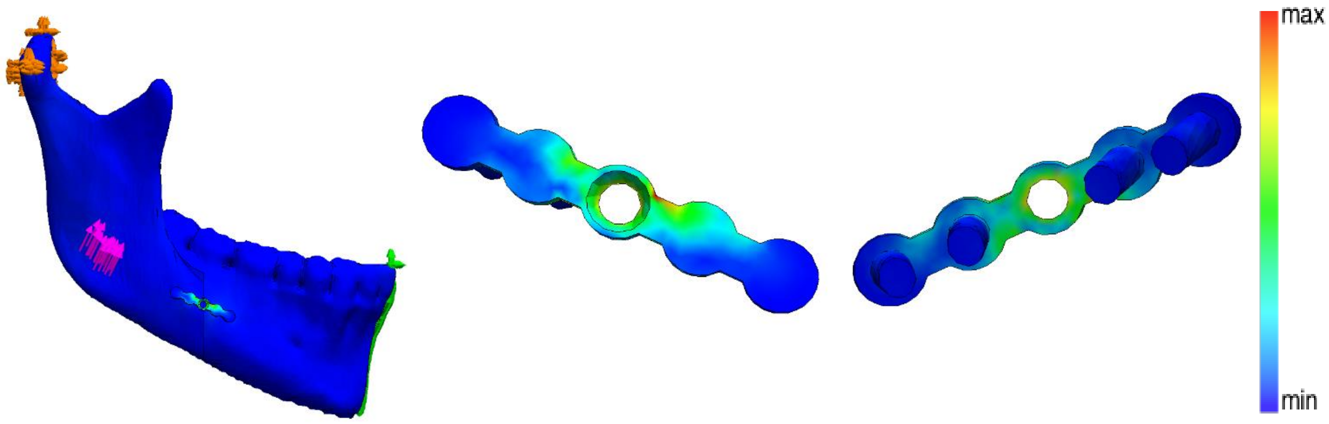
(a) (b) (c)

Figure 6.1 Model 1 (5-hole 1.7mm miniplate) under 50N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



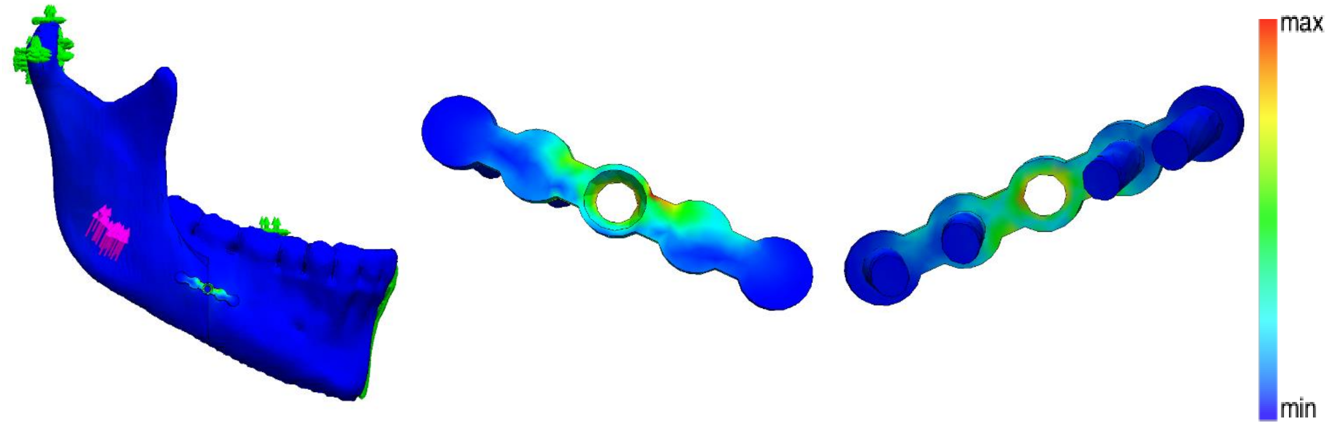
(a) (b) (c)

Figure 6.2 Model 1 (5-hole 1.7mm miniplate) under 75N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



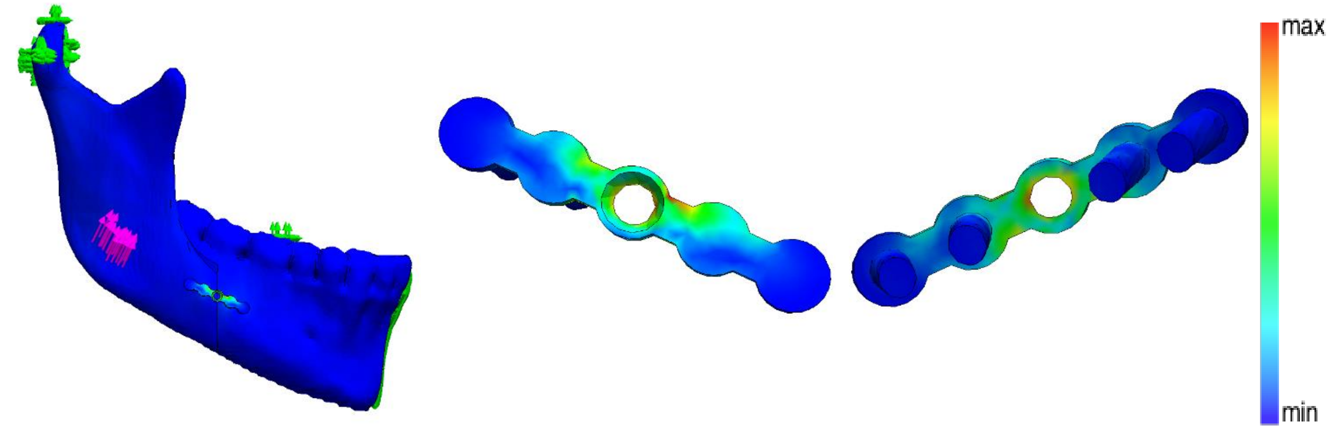
(a) (b) (c)

Figure 6.3: Model 1 (5-hole 1.7mm miniplate) under 100N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



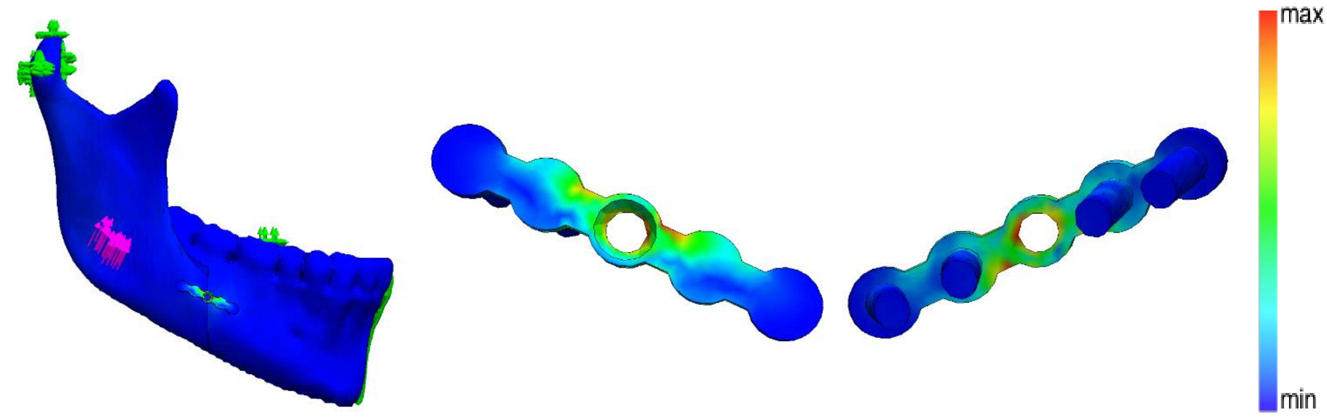
(a) (b) (c)

Figure 6.4 Model 1 (5-hole 1.7mm miniplate) under 100N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



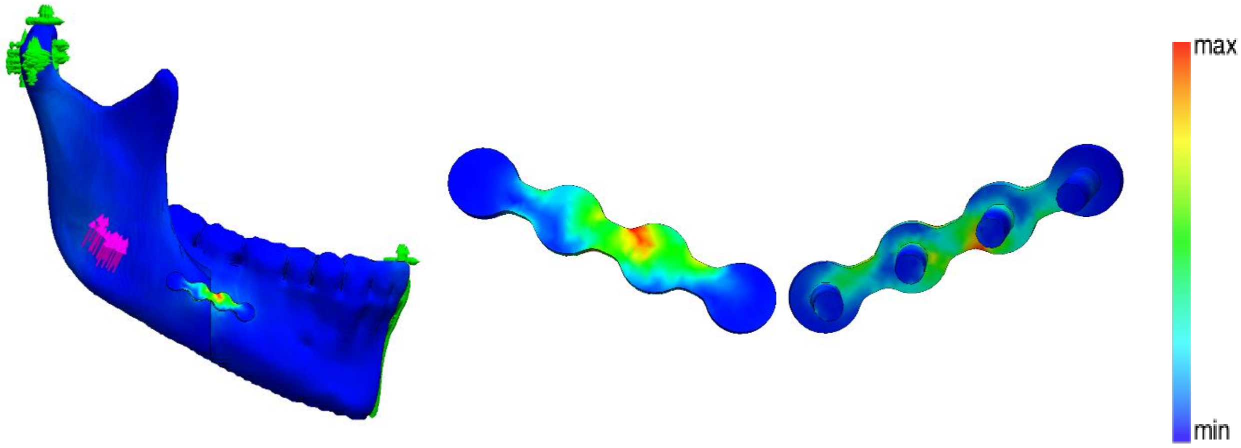
(a) (b) (c)

Figure 6.5 Model 1 (5-hole 1.7mm miniplate) under 200N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



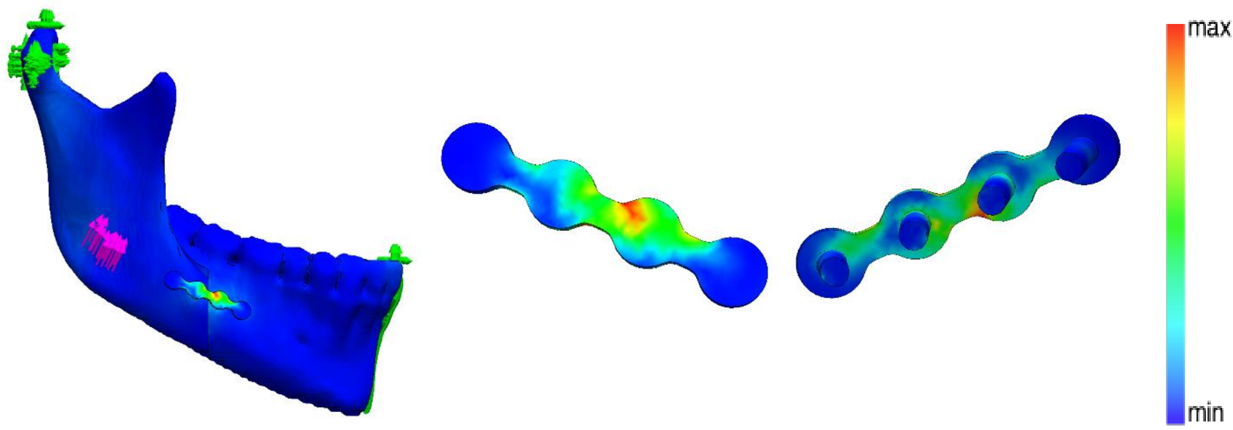
(a) (b) (c)

Figure 6.6 Model 1 (5-hole 1.7mm miniplate) under 300N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



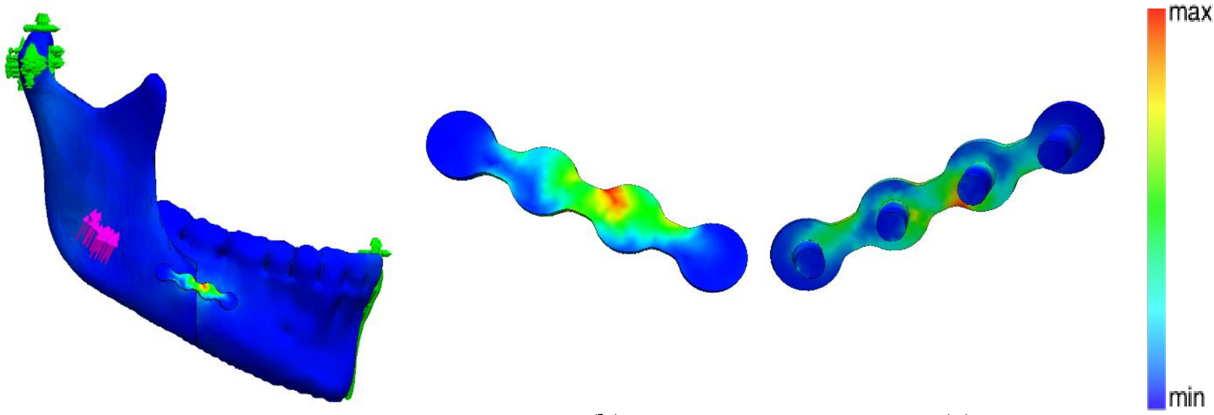
(a) (b) (c)

Figure 6.7 Model 2 (4-hole 2.0mm miniplate with no gap) under 50N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



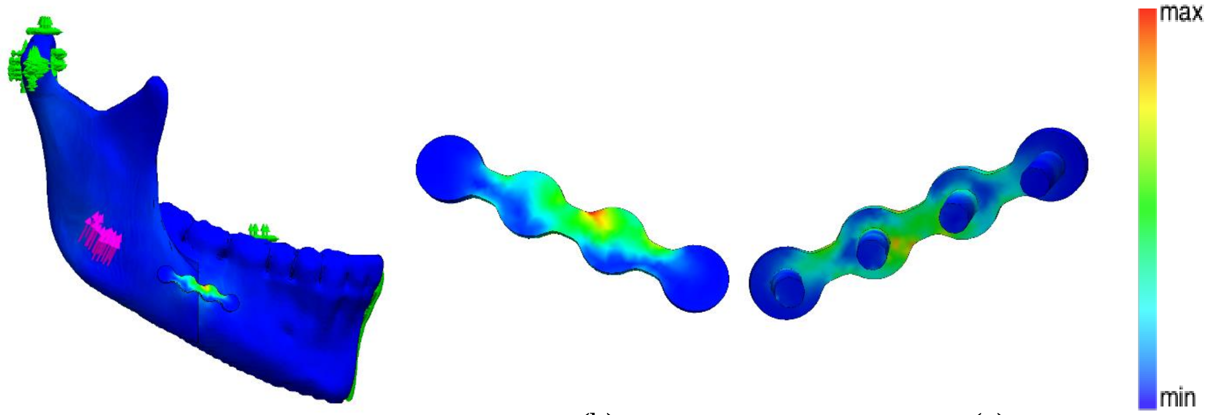
(a) (b) (c)

Figure 6.8 Model 2 (4-hole 2.0mm miniplate with no gap) under 75N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



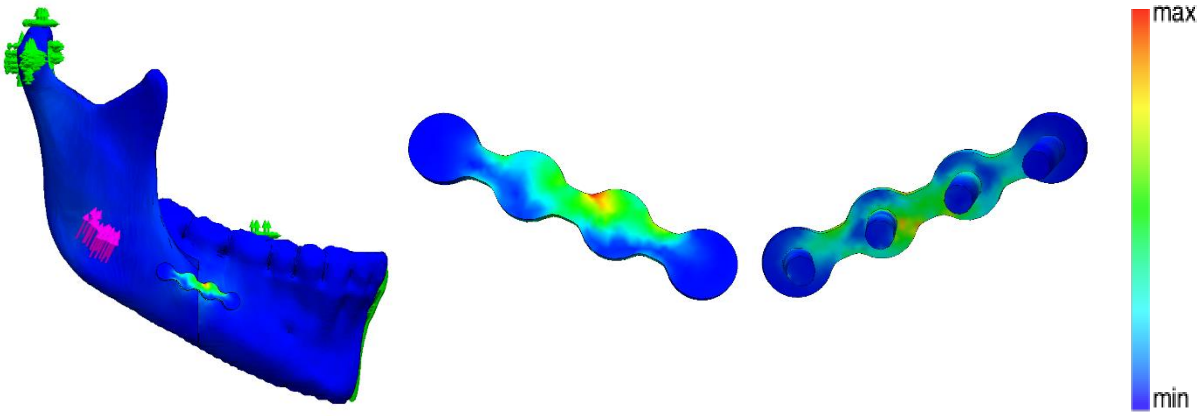
(a) (b) (c)

Figure 6.9 Model 2 (4-hole 2.0mm miniplate with no gap) under 100N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



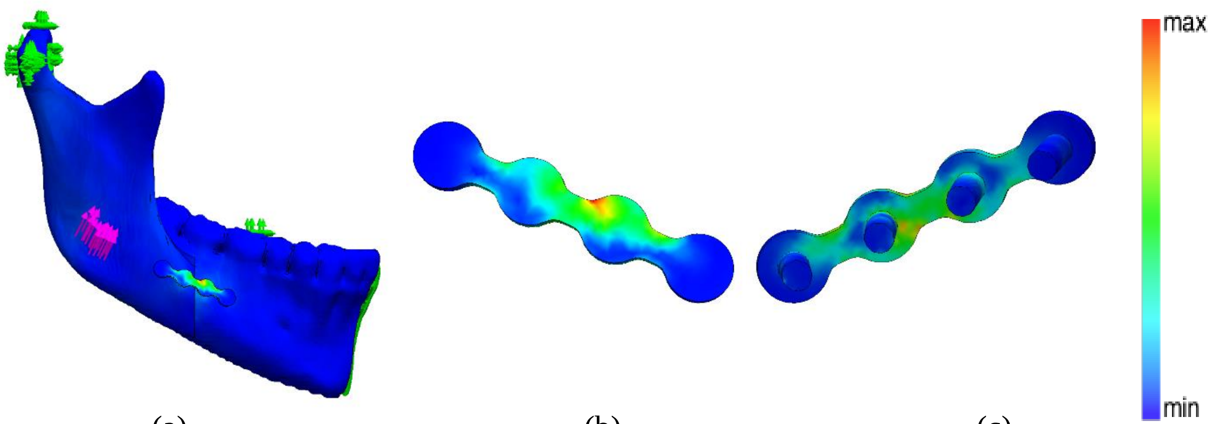
(a) (b) (c)

Figure 6.10 Model 2 (4-hole 2.0mm miniplate with no gap) under 100N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



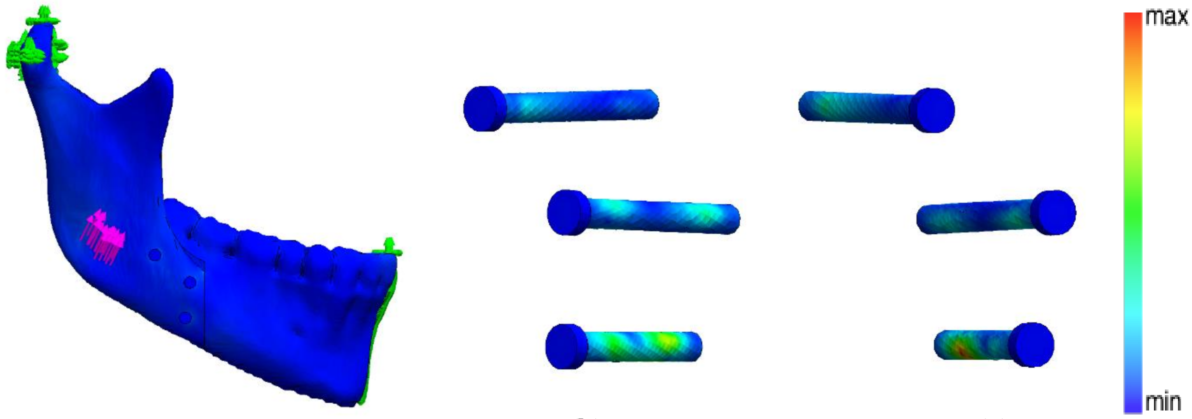
(a) (b) (c)

Figure 6.11 Model 2 (4-hole 2.0mm miniplate with no gap) under 200N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



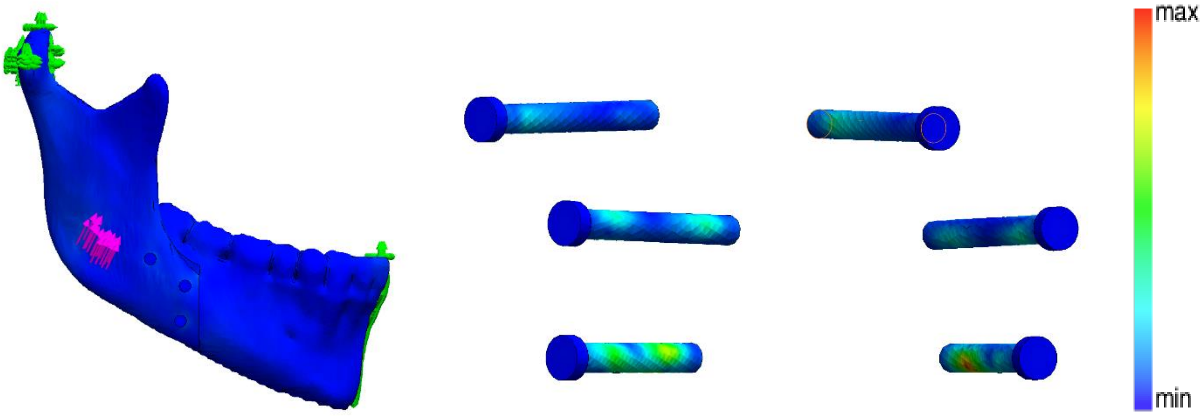
(a) (b) (c)

Figure 6.12 Model 2 (4-hole 2.0mm miniplate with no gap) under 300N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



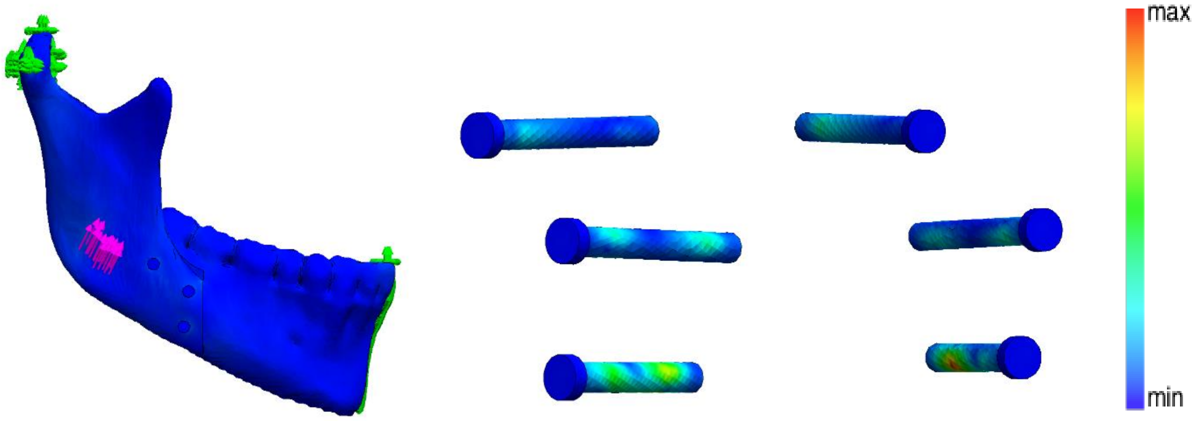
(a) (b) (c)

Figure 6.13 Model 3 under 50N loading and incisal restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



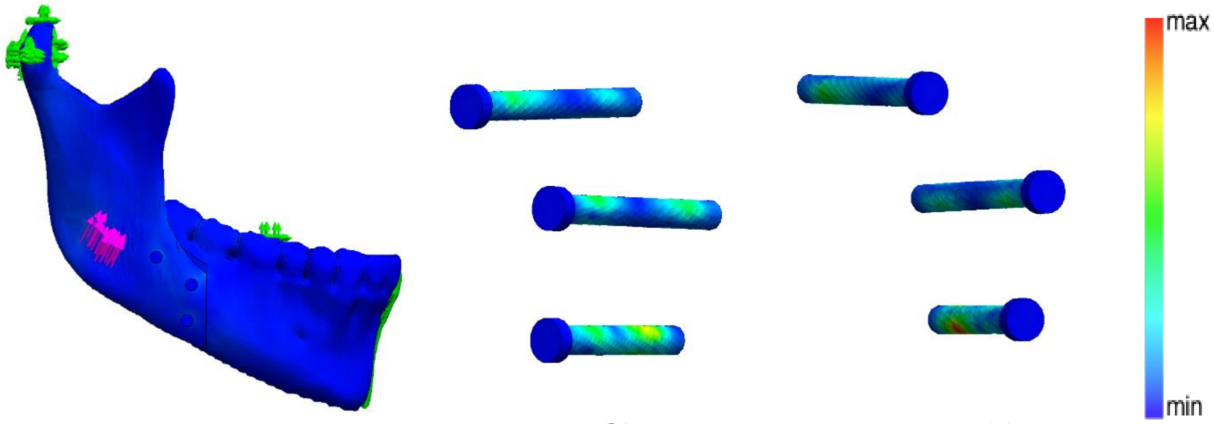
(a) (b) (c)

Figure 6.14 Model 3 under 75N loading and incisal restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



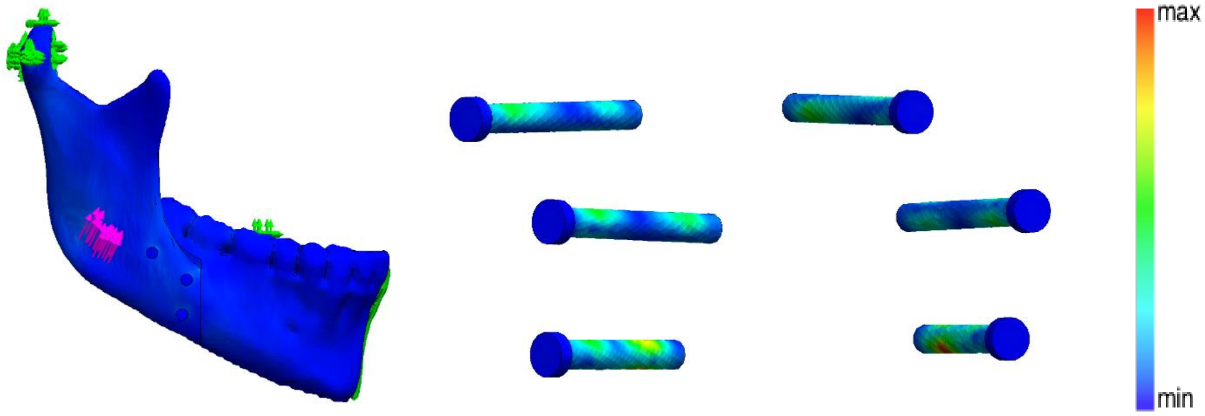
(a) (b) (c)

Figure 6.15 Model 3 under 100N loading and incisal restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



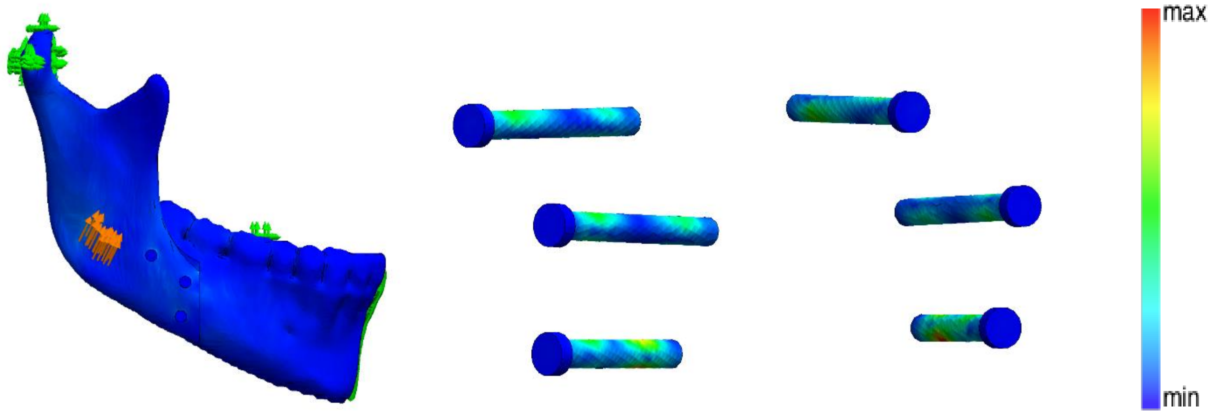
(a) (b) (c)

Figure 6.16 Model 3 under 100N loading and molar restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



(a) (b) (c)

Figure 6.17 Model 3 under 200N loading and molar restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



(a) (b) (c)

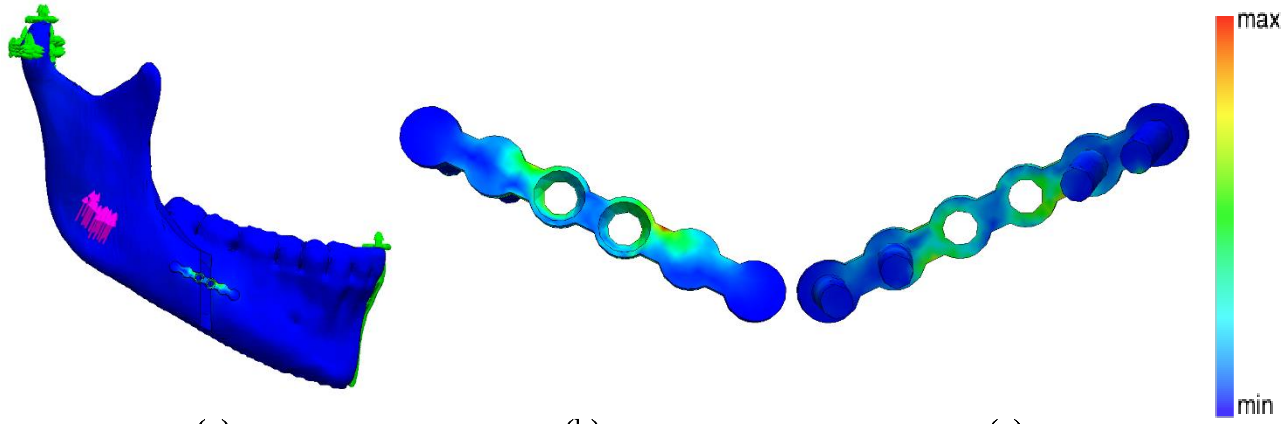
Figure 6.18 Model 3 under 300N loading and molar restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation

6.3.2 Advancement 3mm – Stress distribution in models and fixation techniques

Model 4 incorporated a 6-hole 1.7mm miniplate with the 3rd and 4th hole acted as the connector (Fig 6.20 – 6.25). Stresses in this model are generally distributed between the 2nd and 5th hole of the miniplate and maximum stress concentrated on the superior margin between the 4th and 5th hole/screw, and of the fixation.

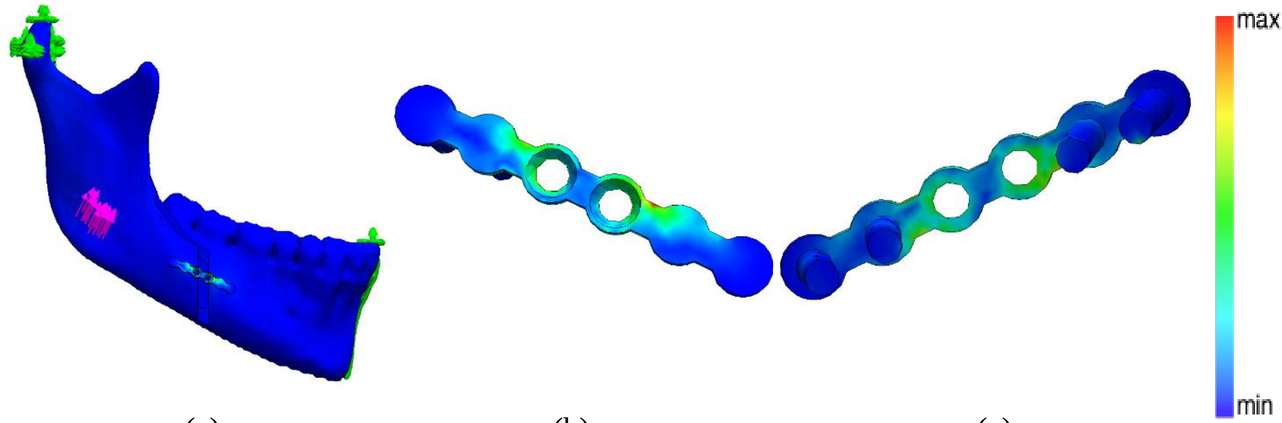
Stresses in Model 5 also distributes in the connector region of the 2.0mm miniplate with maximum stress concentrating in the superior – proximal margin of the 3rd hole and inferior distal margin of the 2nd hole (Fig. 6.26 – 6.31).

Model 6 stress distribution is similar to that of Model 3 (Fig. 6.32 – 6.37). More stress distribution is seen in the inferior – distal screw, with maximum stress located at the lingual cortical – cancellous bone junction.



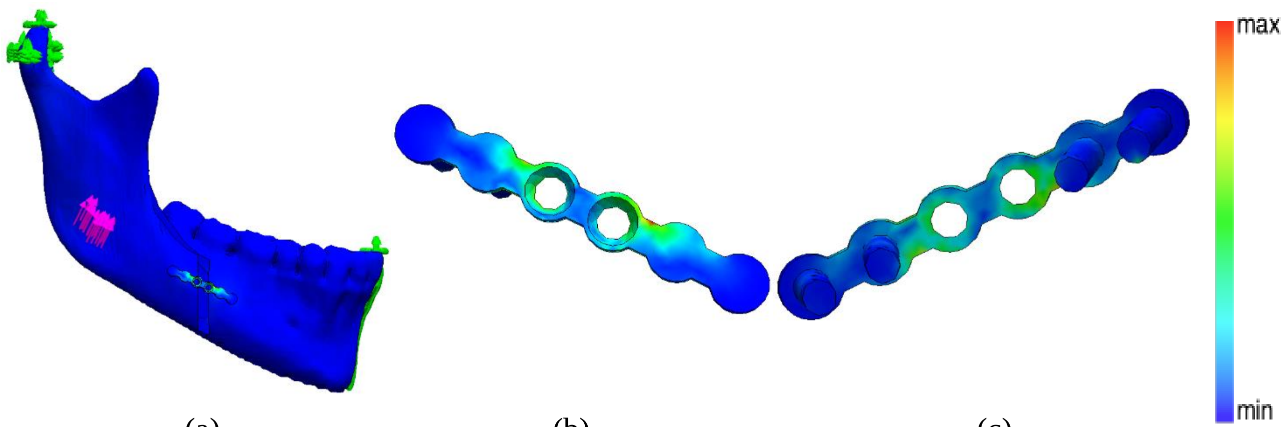
(a) (b) (c)

Figure 6.19 Model 4 (6-hole 1.7mm miniplate) under 50N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



(a) (b) (c)

Figure 6.20 Model 4 (6-hole 1.7mm miniplate) under 75N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



(a) (b) (c)

Figure 6.21 Model 4 (6-hole 1.7mm miniplate) under 100N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

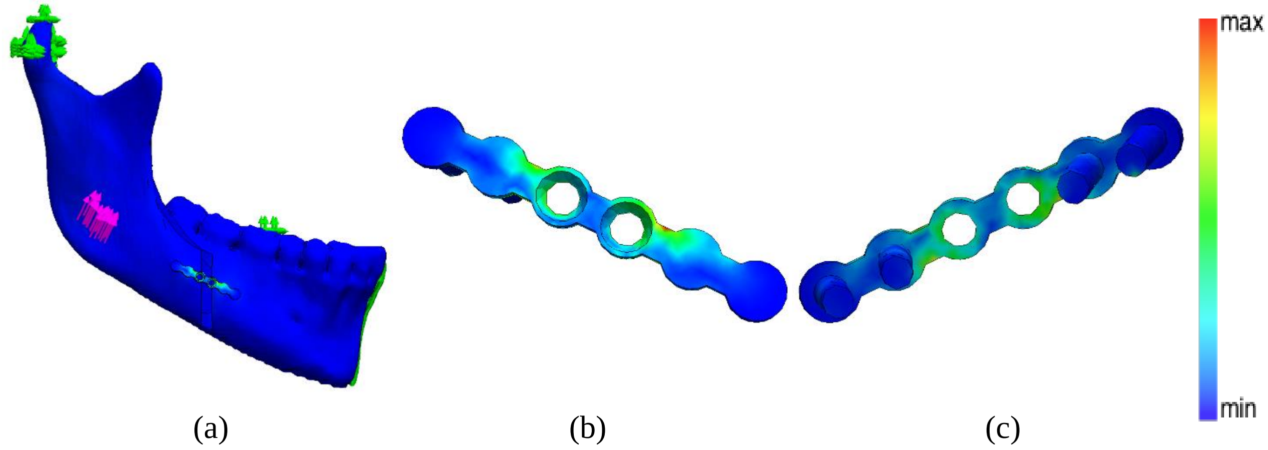


Figure 6.22 Model 4 (6-hole 1.7mm miniplate) under 100N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

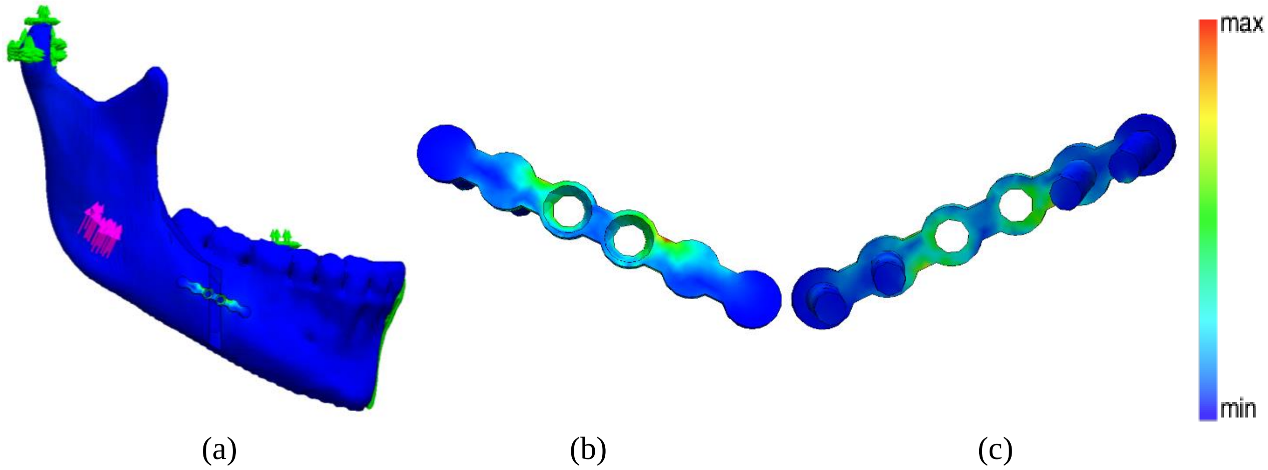


Figure 6.23 Model 4 (6-hole 1.7mm miniplate) under 200N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

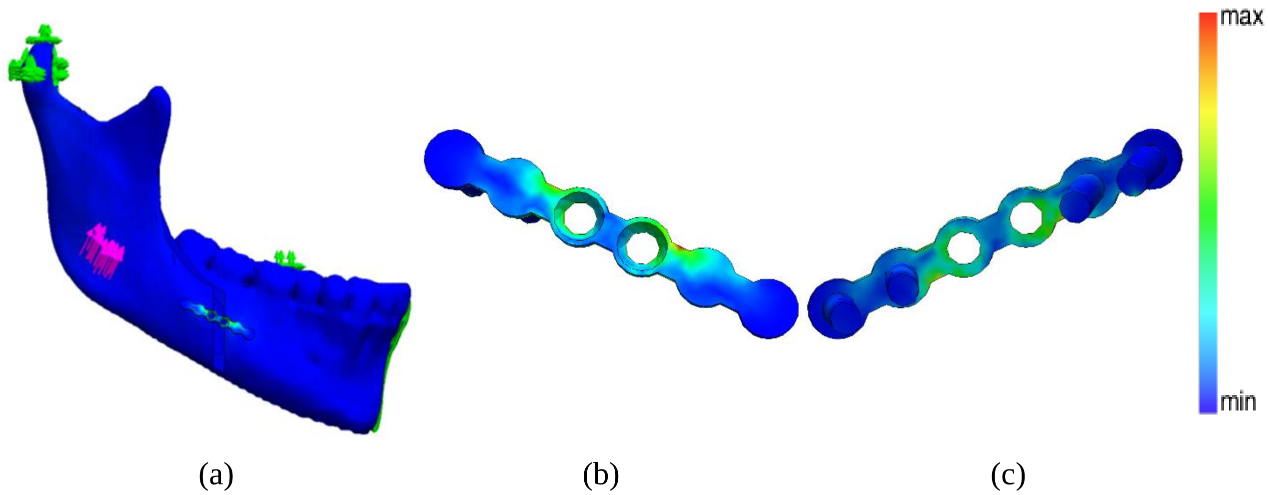
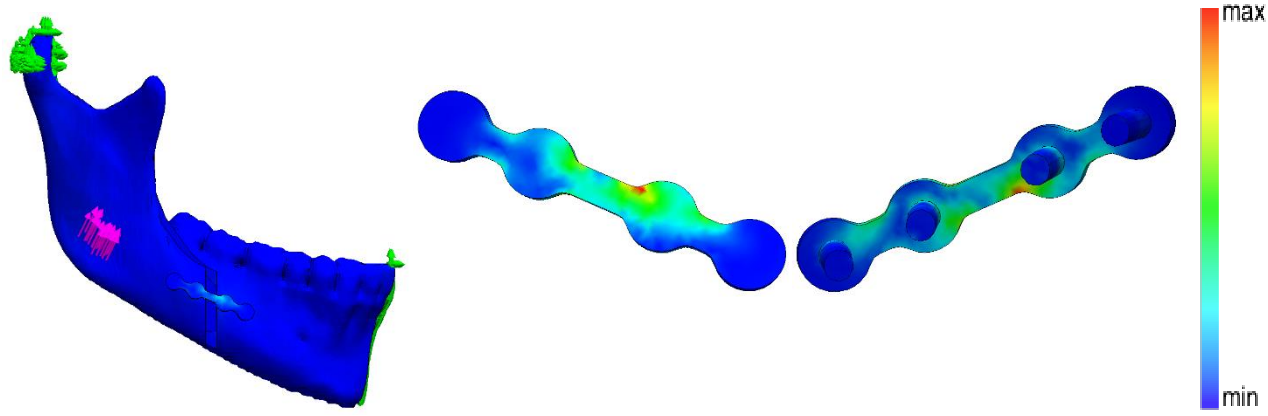
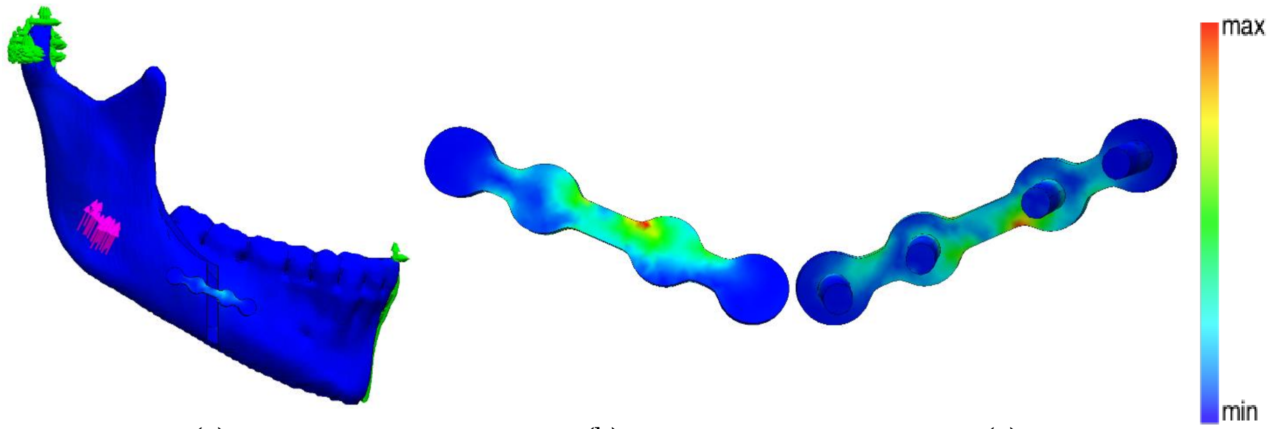


Figure 6.24 Model 4 (6-hole 1.7mm miniplate) under 300N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



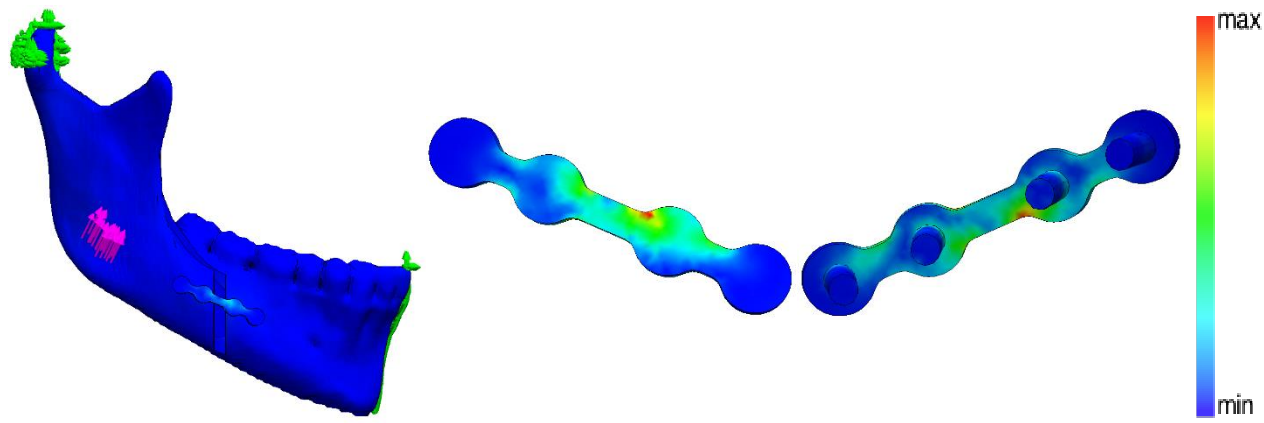
(a) (b) (c)

Figure 6.25 Model 5 (2.0mm miniplate with 4.4mm gap) under 50N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



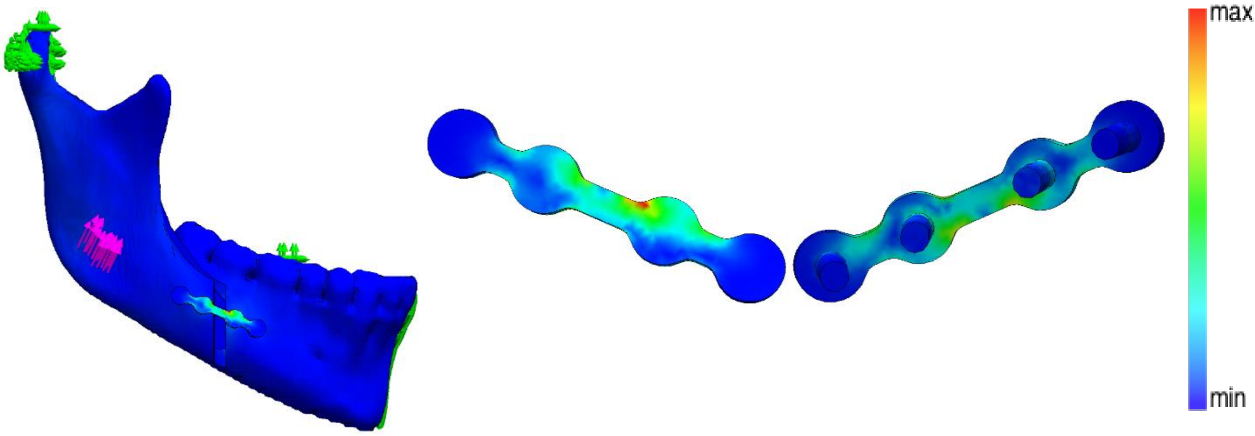
(a) (b) (c)

Figure 6.26 Model 5 (2.0mm miniplate with 4.4mm gap) under 75N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



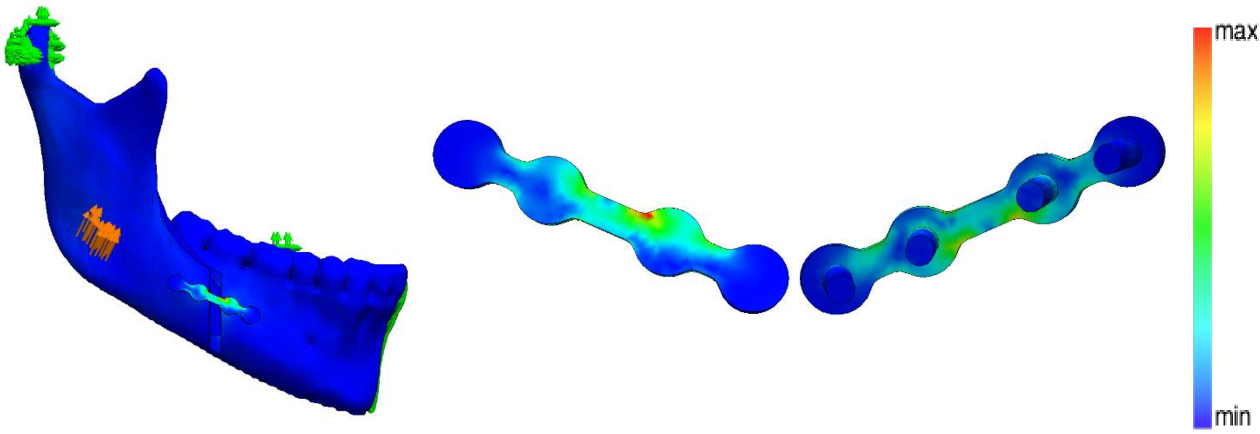
(a) (b) (c)

Figure 6.27 Model 5 (2.0mm miniplate with 4.4mm gap) under 100N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



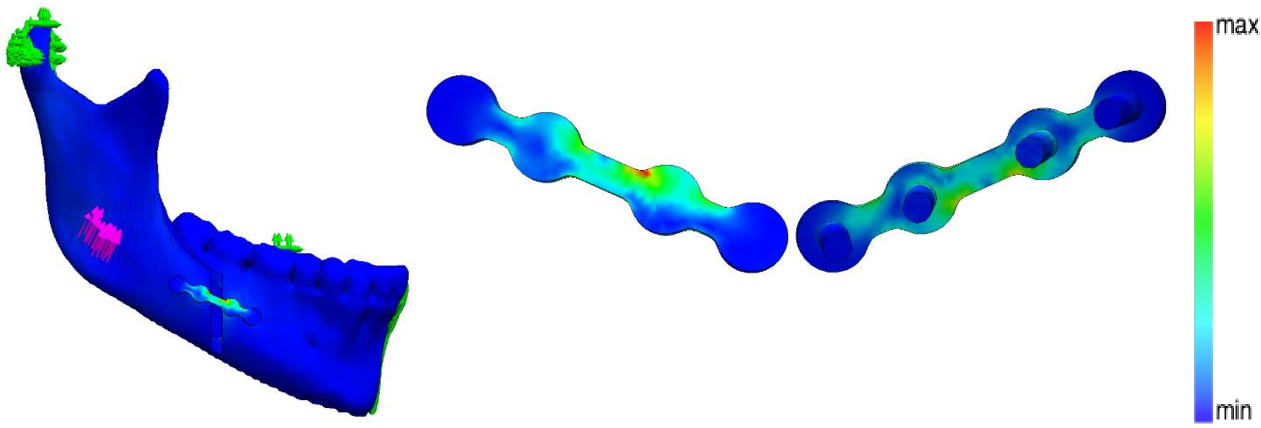
(a) (b) (c)

Figure 6.28 Model 5 (2.0mm miniplate with 4.4mm gap) under 100N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



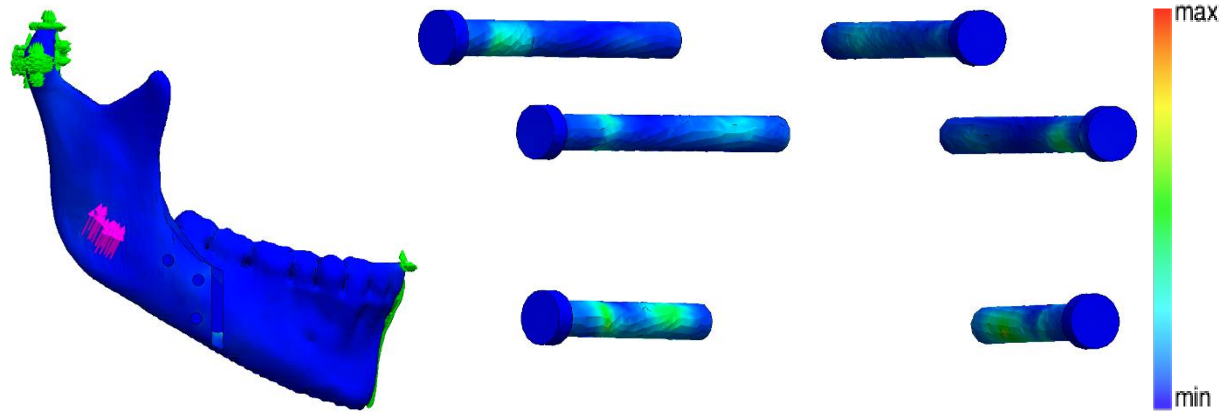
(a) (b) (c)

Figure 6.29 Model 5 (2.0mm miniplate with 4.4mm gap) under 200N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



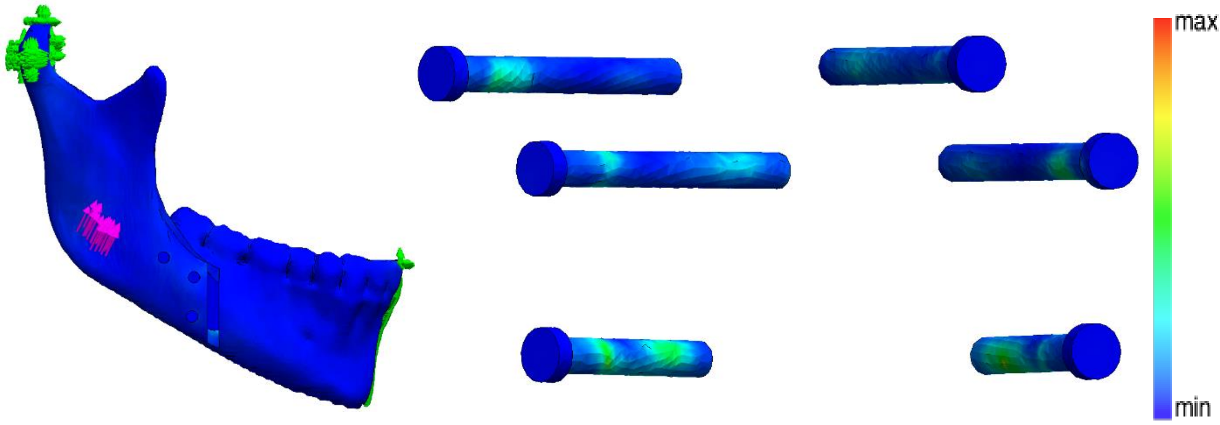
(a) (b) (c)

Figure 6.30 Model 5 (2.0mm miniplate with 4.4mm gap) under 300N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



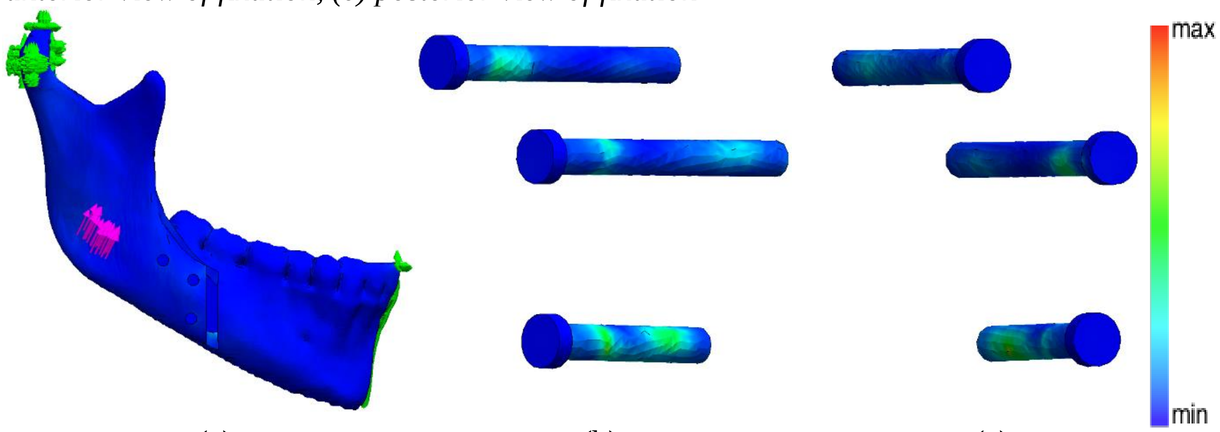
(a) (b) (c)

Figure 6.31 Model 6 under 50N loading and incisal restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



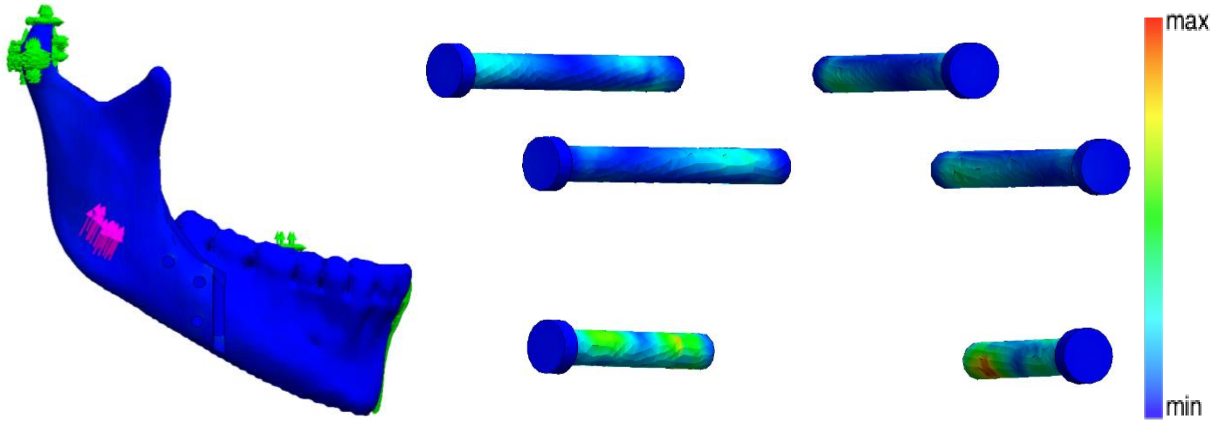
(a) (b) (c)

Figure 6.32 Model 6 under 75N loading and incisal restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



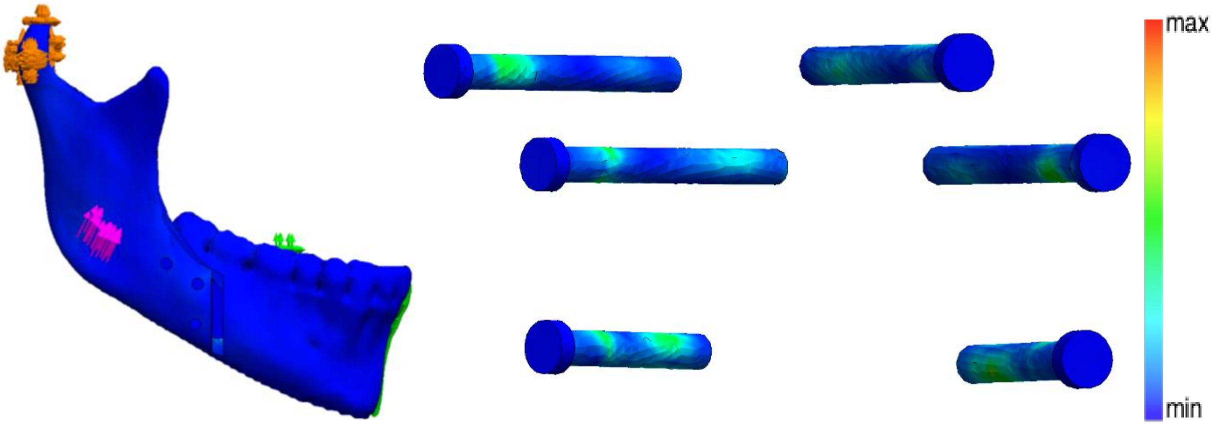
(a) (b) (c)

Figure 6.33 Model 6 under 100N loading and incisal restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



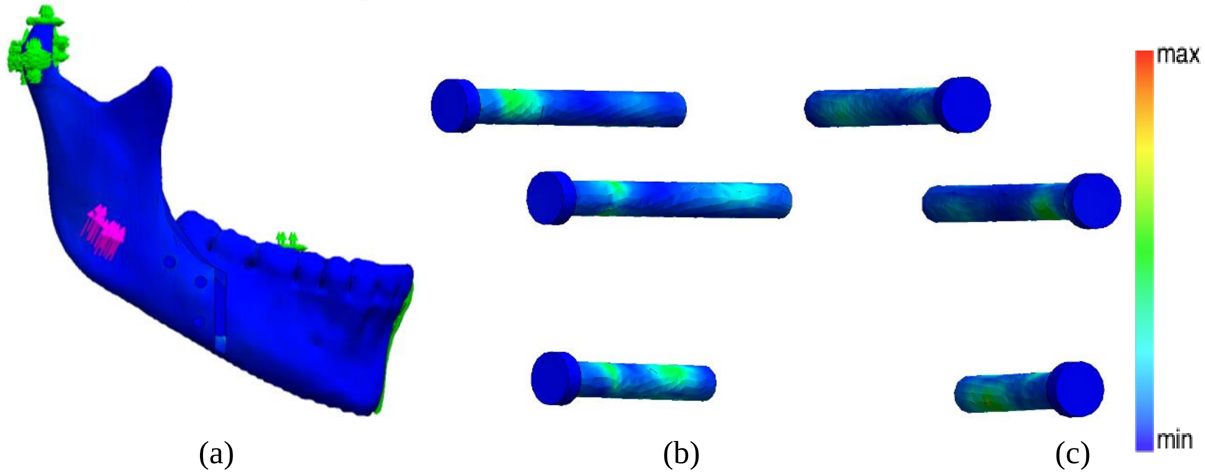
(a) (b) (c)

Figure 6.34 Model 6 under 100N loading and molar restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



(a) (b) (c)

Figure 6.35 Model 6 under 200N loading and molar restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation



(a) (b) (c)

Figure 6.36 Model 6 under 300N loading and molar restraint. (a) buccal view of model, (b) anterior view of fixation, (c) posterior view of fixation

6.3.3 Advancement 7mm – Stress distribution in models and fixation techniques

Stresses in Model 7 distributes mainly around the 2nd, 3rd, 5th and 6th hole. The maximum stress is seen on the superior - distal margin of the 5th hole. High stresses are also evident in the inferior margin between the 2nd and 3rd hole of the 1.7mm miniplate fixation.

Model 8 show stress distributions closer to the 2nd and 3rd screws of the 2.0mm miniplate fixation. It is also evident that stress is distributed along the superior region of the connector on the buccal surface and along the inferior region lingually. Maximum stress is located at the inferior distal surface of the 2nd hole with high stress region also evident on the superior – proximal region of the 3rd hole.

Model 9 stress distribution is similar in previous mandibular movements. More stress is distributed in the inferior-distal screw. Maximum stresses in this screw is located on the inferior region cortical-cancellous bone junction.

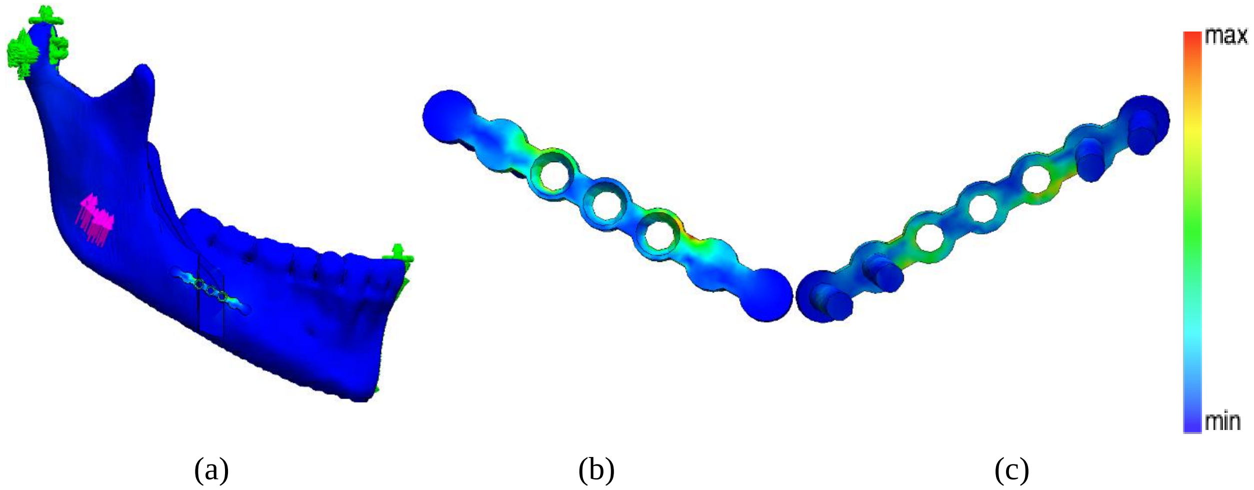


Figure 6.37 Model 7(7-hole 1.7mm miniplate) under 50N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

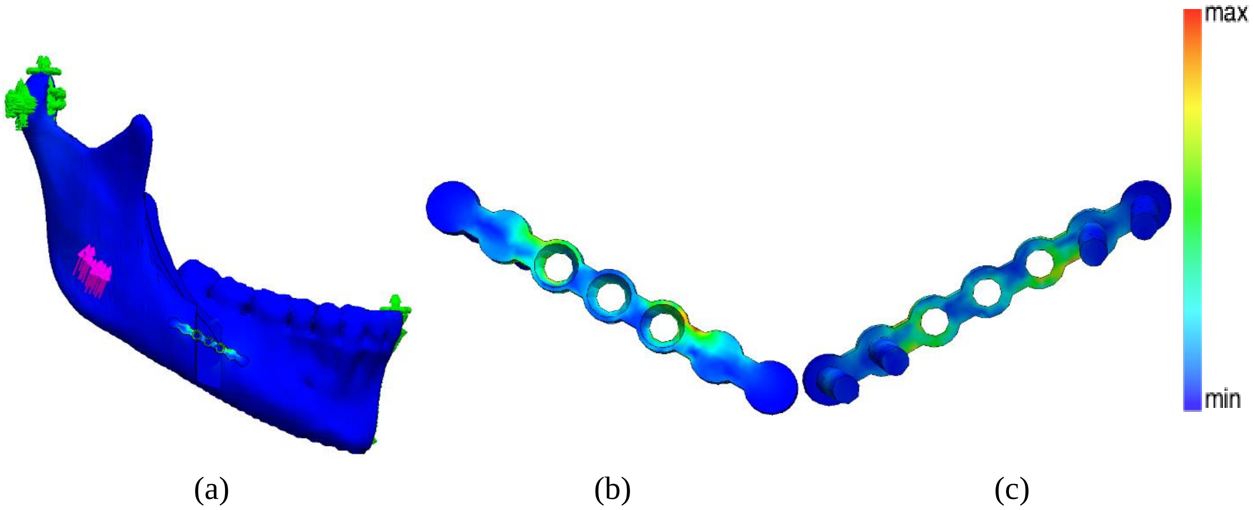


Figure 6.38 Model 7 (7-hole 1.7mm miniplate) under 75N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

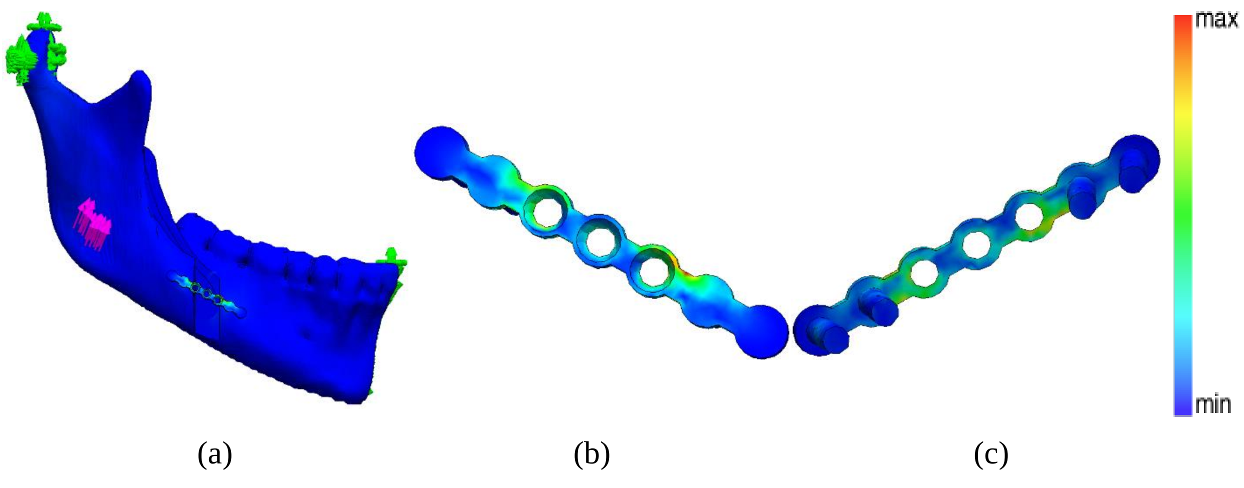
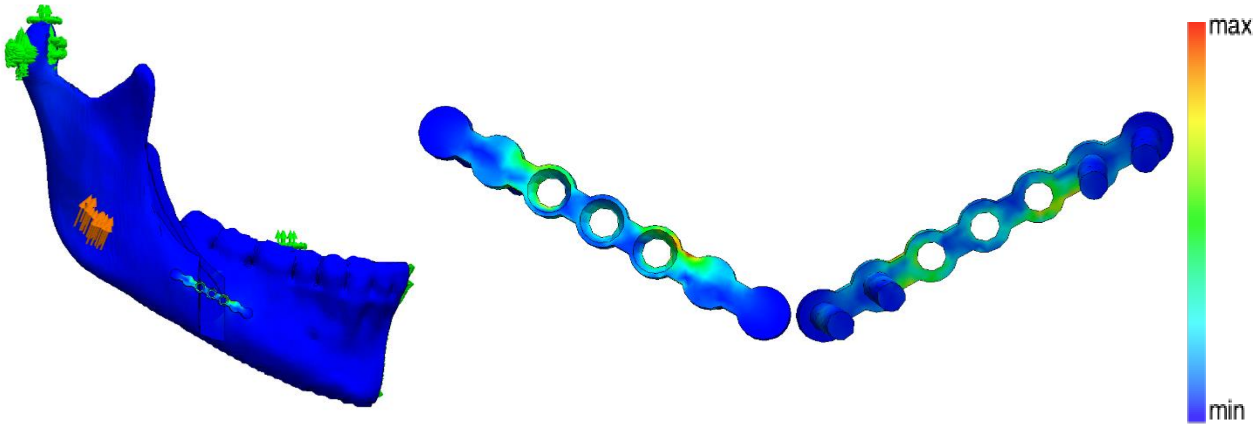
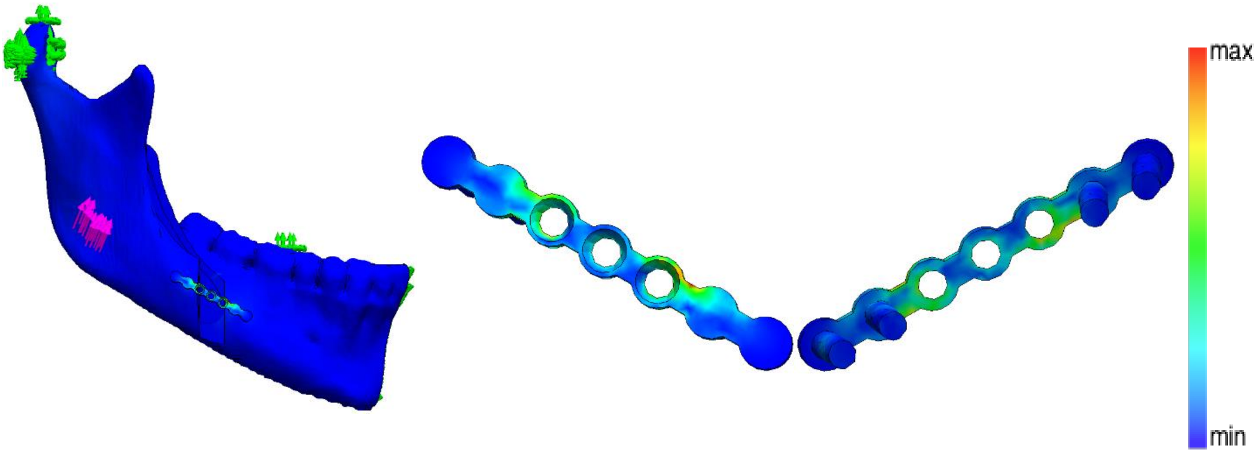


Figure 6.39 Model 7(7-hole 1.7mm miniplate) under 100N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



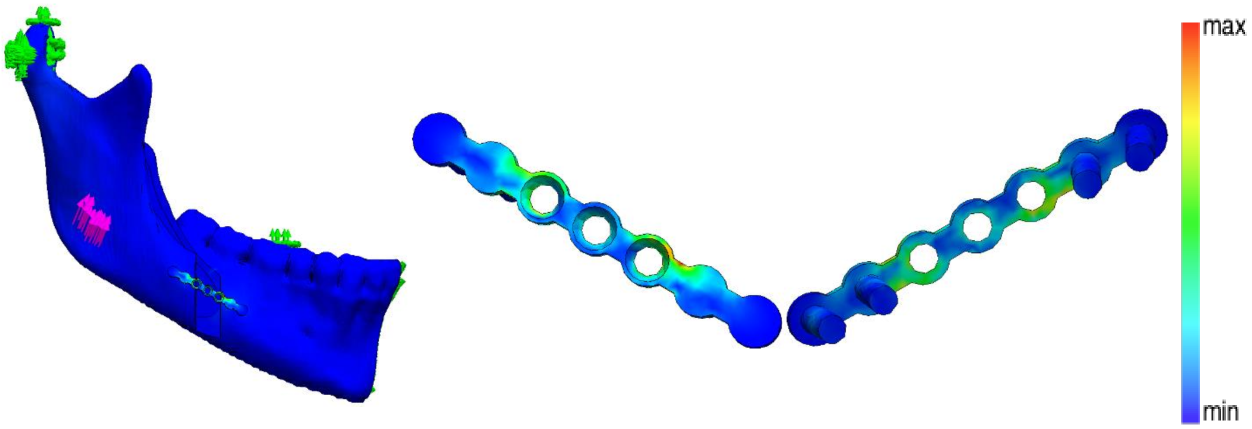
(a) (b) (c)

Figure 6.40 Model 7 (7-hole 1.7mm miniplate) under 100N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



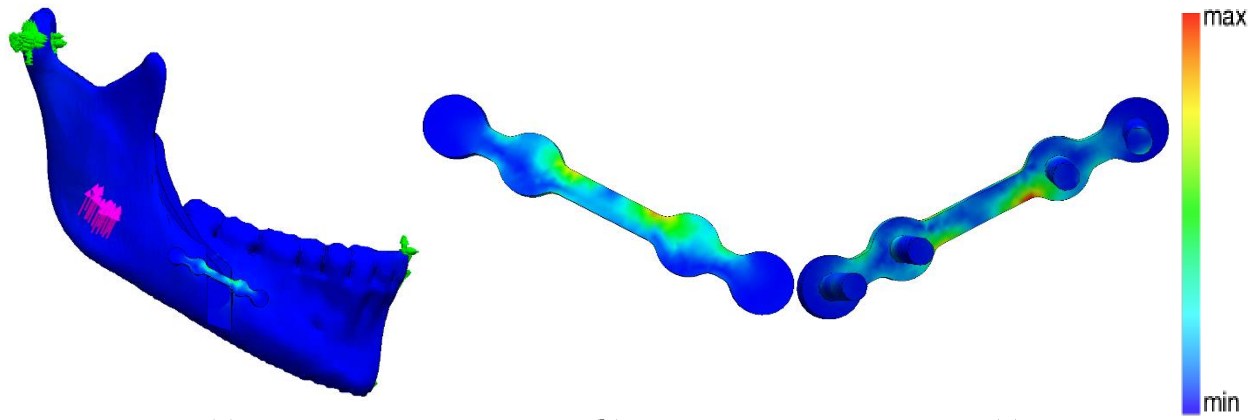
(a) (b) (c)

Figure 6.41 Model 7 (7-hole 1.7mm miniplate) under 200N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



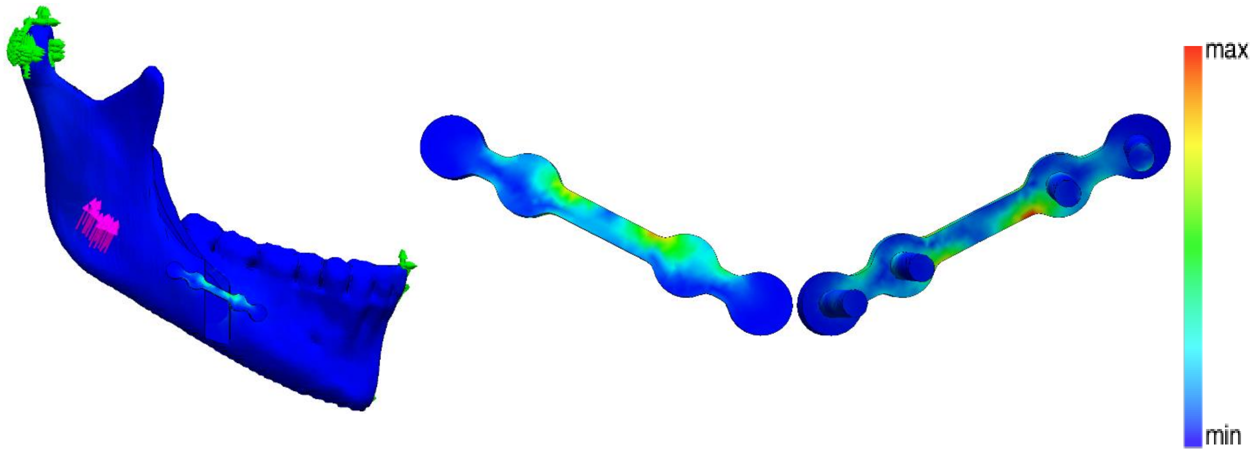
(a) (b) (c)

Figure 6.42 Model 7 (7-hole 1.7mm miniplate) under 300N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



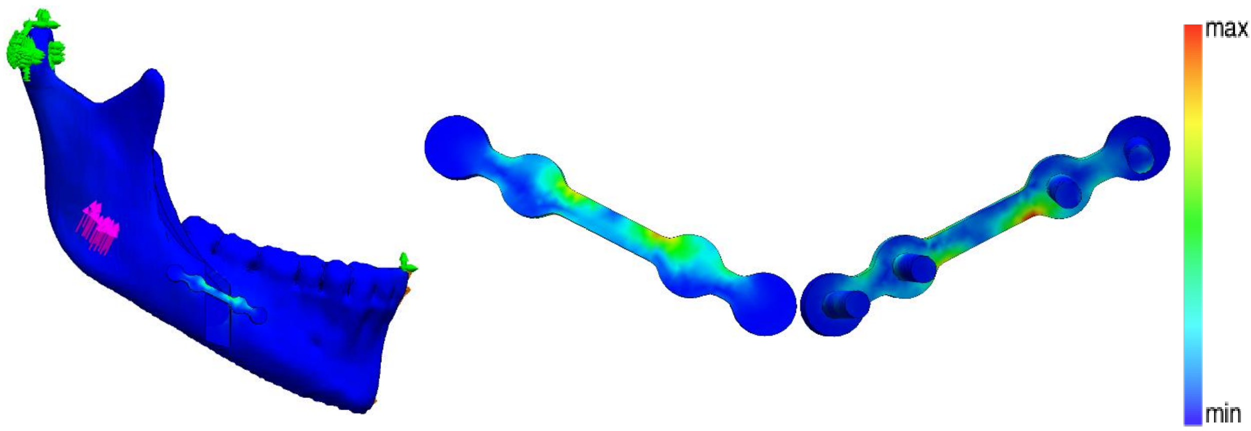
(a) (b) (c)

Figure 6.43 Model 8 (2.0mm miniplate with 8mm gap) under 50N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



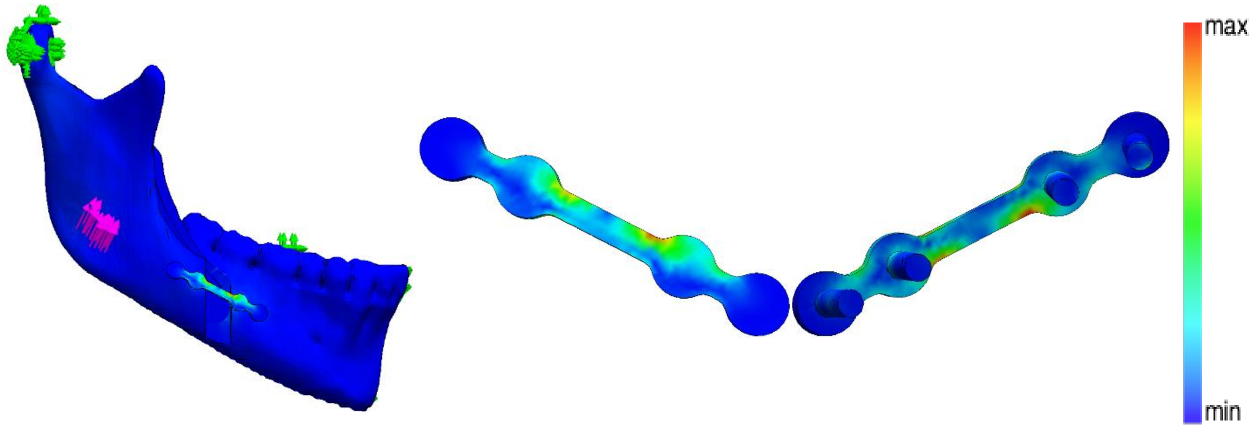
(a) (b) (c)

Figure 6.44 Model 8 (2.0mm miniplate with 8mm gap) under 75N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



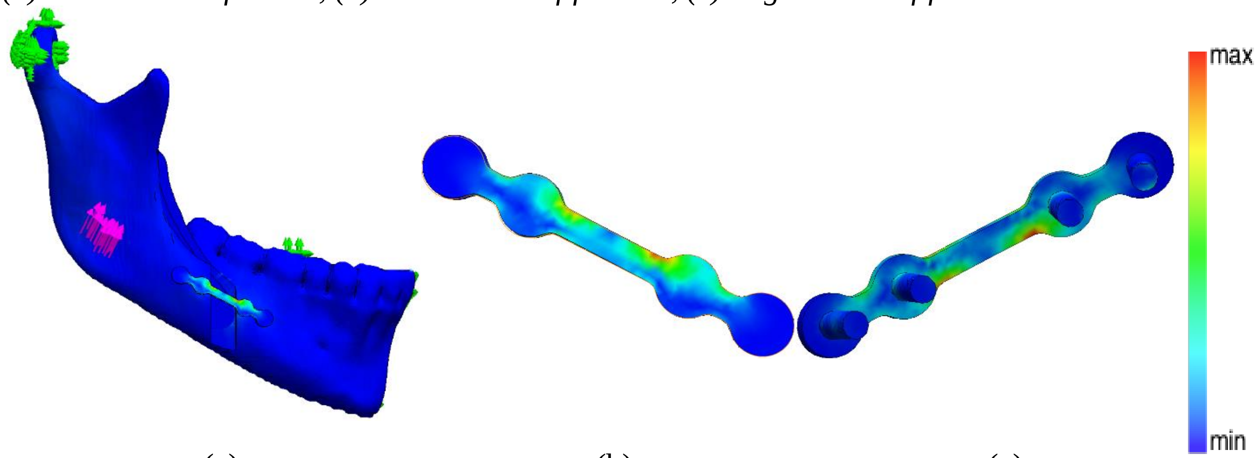
(a) (b) (c)

Figure 6.45 Model 8 (2.0mm miniplate with 8mm gap) under 100N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



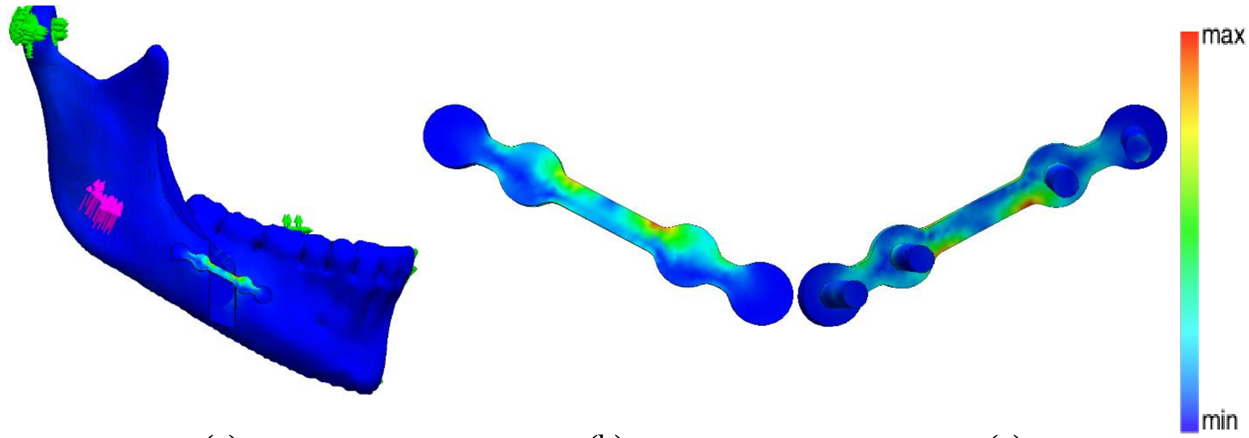
(a) (b) (c)

Figure 6.46 Model 8 (2.0mm miniplate with 8mm gap) under 100N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



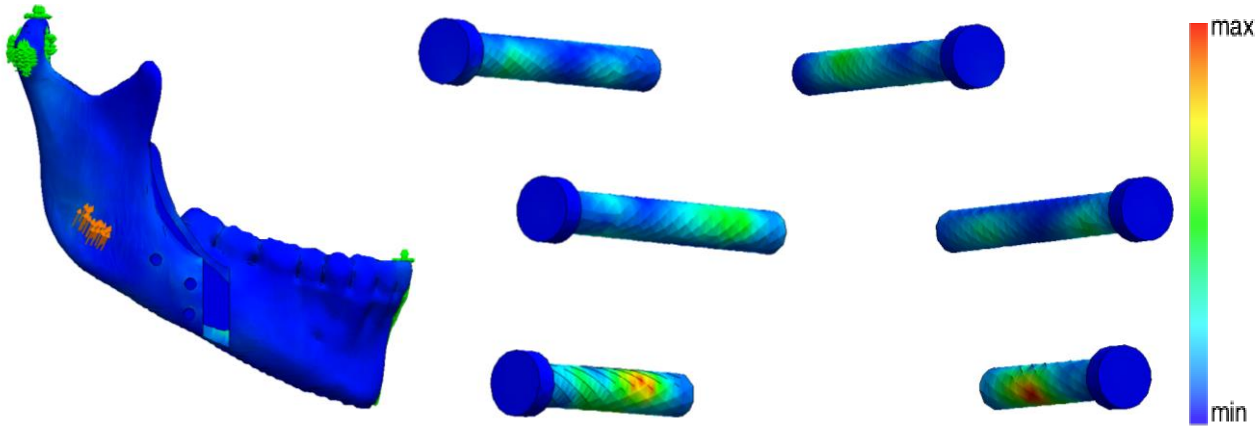
(a) (b) (c)

Figure 6.47 Model 8 (2.0mm miniplate with 8mm gap) under 200N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



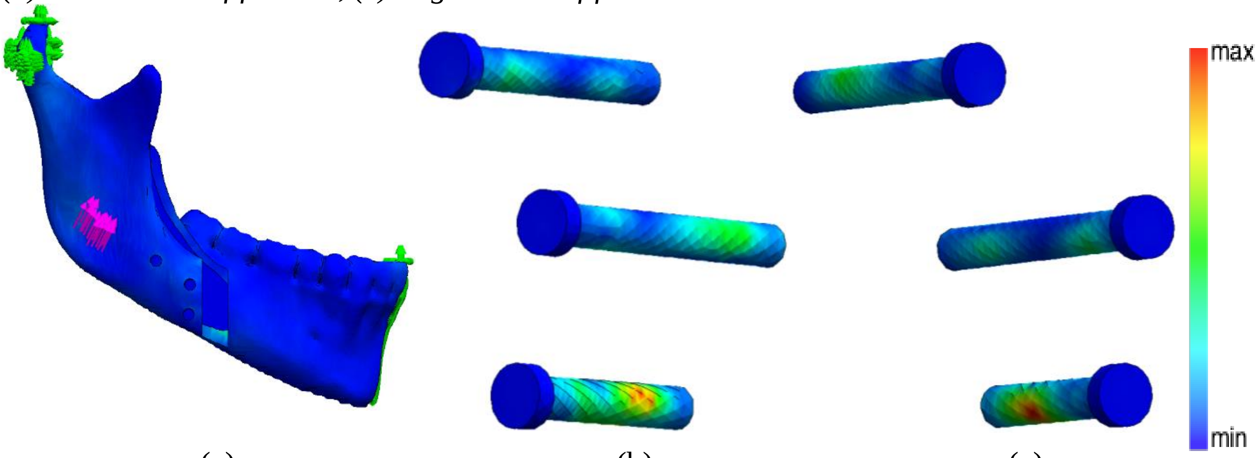
(a) (b) (c)

Figure 6.48 Model 8 (2.0mm miniplate with 8mm gap) under 300N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



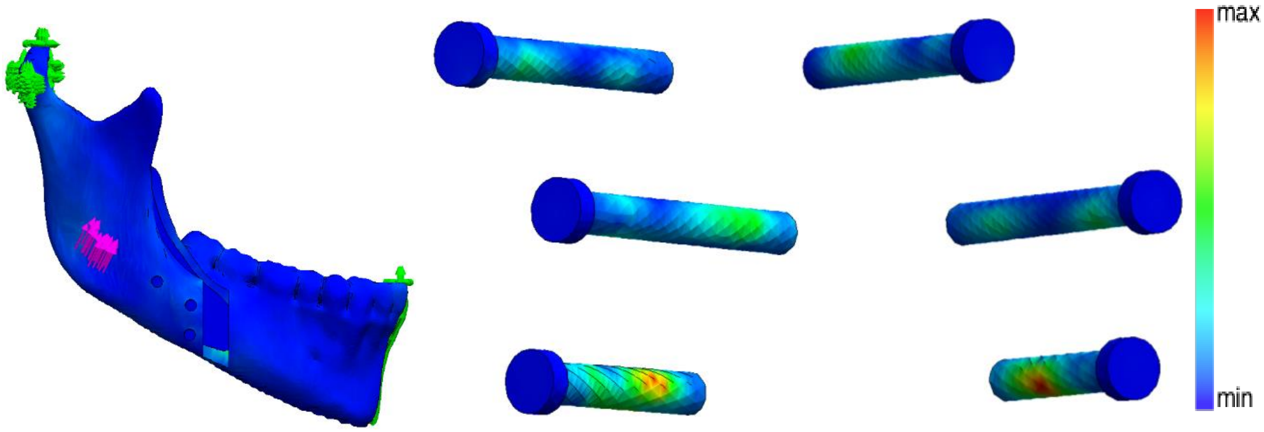
(a) (b) (c)

Figure 6.49 Model 9 under 50N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



(a) (b) (c)

Figure 6.50 Model 9 under 75N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation



(a) (b) (c)

Figure 6.51 Model 9 under 100N loading and incisal restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

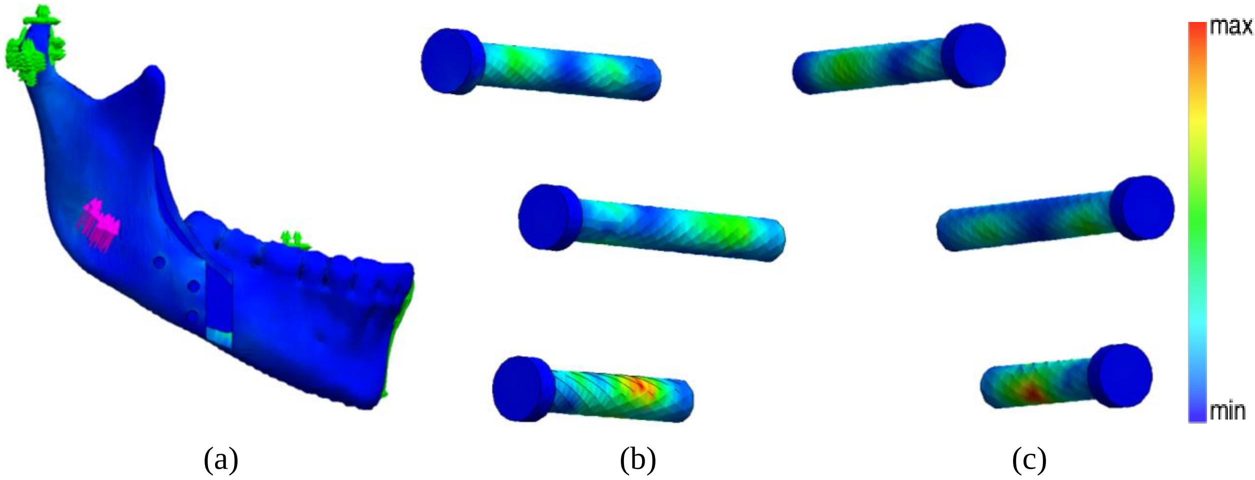


Figure 6.52 Model 9 under 100N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

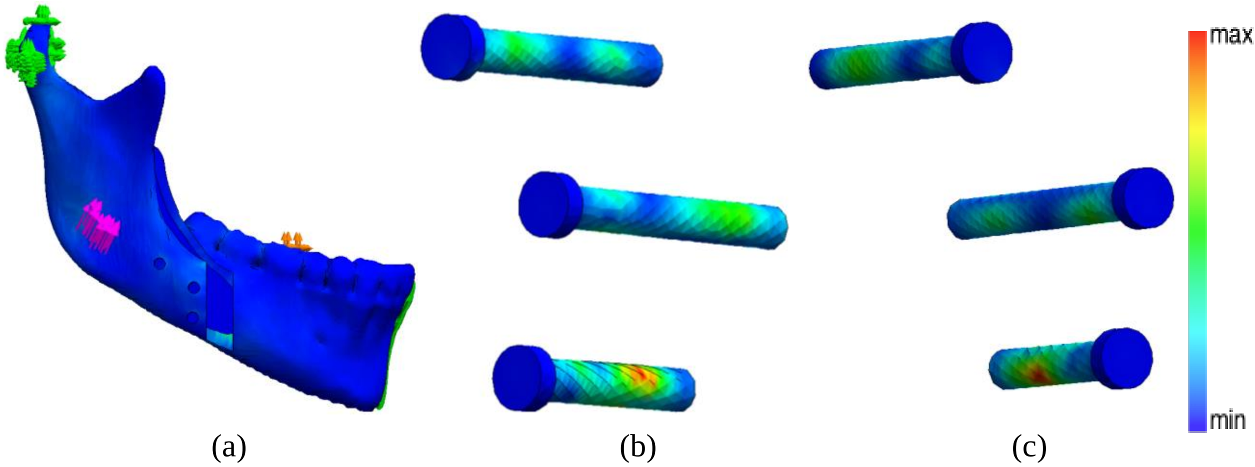


Figure 6.53 Model 9 under 200N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

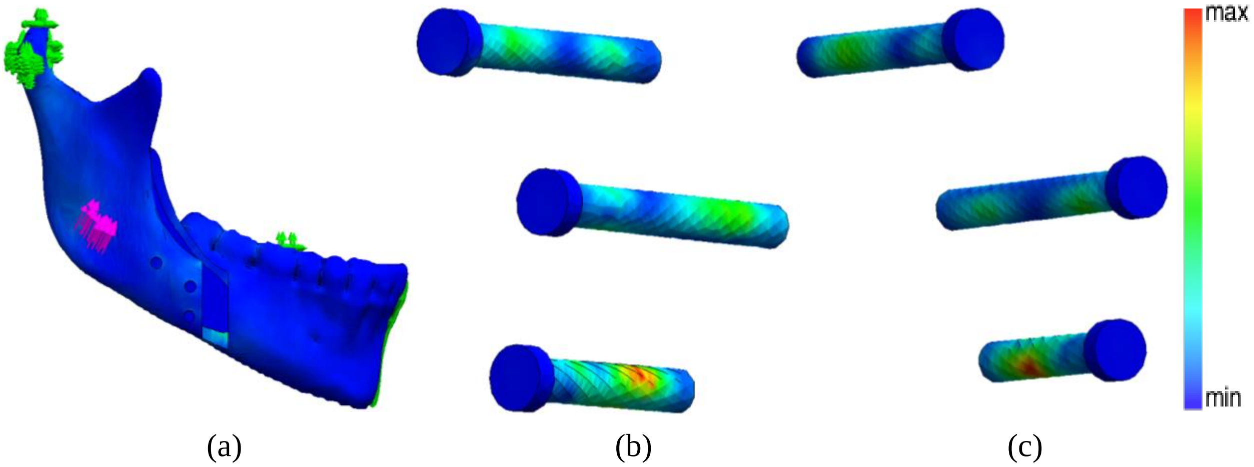


Figure 6.54 Model 9 under 300N loading and molar restraint. (a) buccal view of model, (b) buccal view of fixation, (c) lingual view of fixation

6.3.4 Comparison of Stress in Fixation Techniques

Maximum Von Mises stress observed in all three fixation techniques varied with force magnitude, and mandibular osteotomy distance. Highest stress among all three fixation techniques is seen in the 1.7mm miniplate, while the least stress recorded was seen in the 2.0mm Bi-cortical screws fixation. An exception is seen in the 3mm mandibular setback where the 2.0mm miniplate recorded less stress in the region of forces 50N – 100N with incisal restraint on the model.

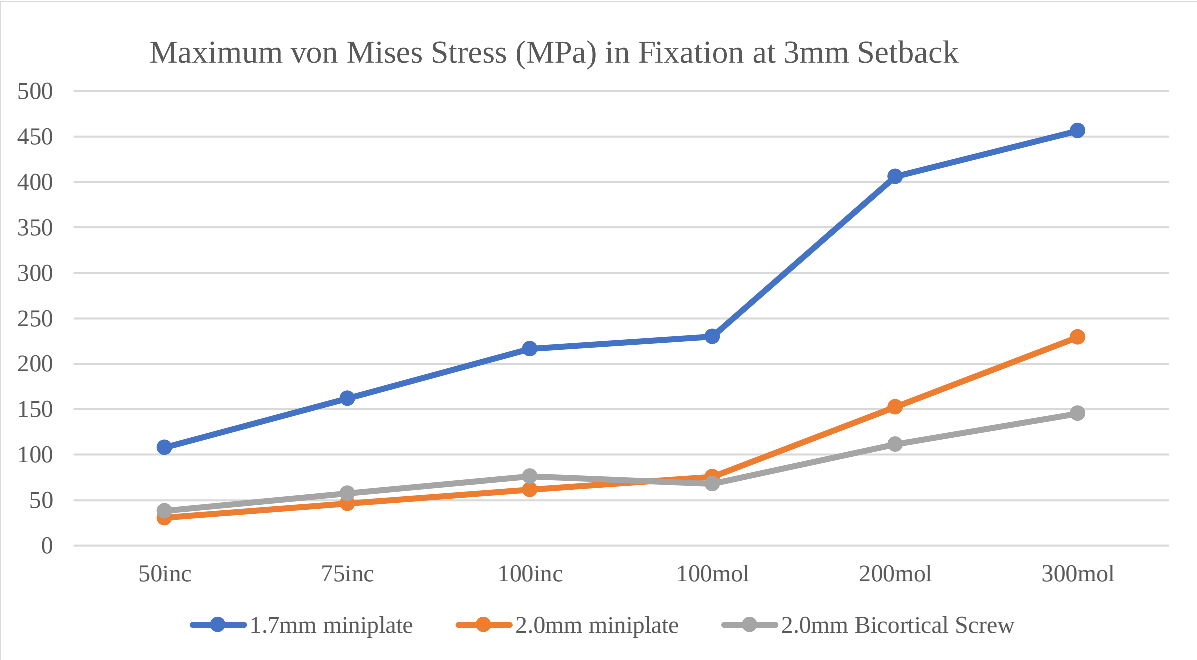


Figure 6.55 Stress in fixation techniques at 3mm Setback

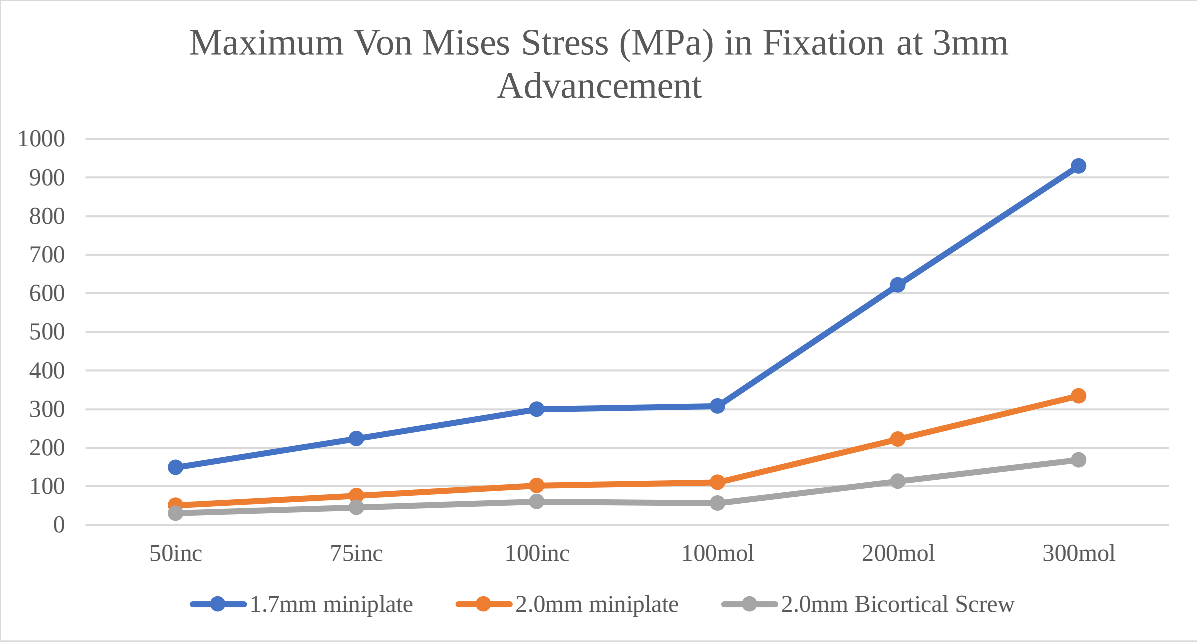


Figure 6.56 Stress in fixation techniques at 3mm Advancement

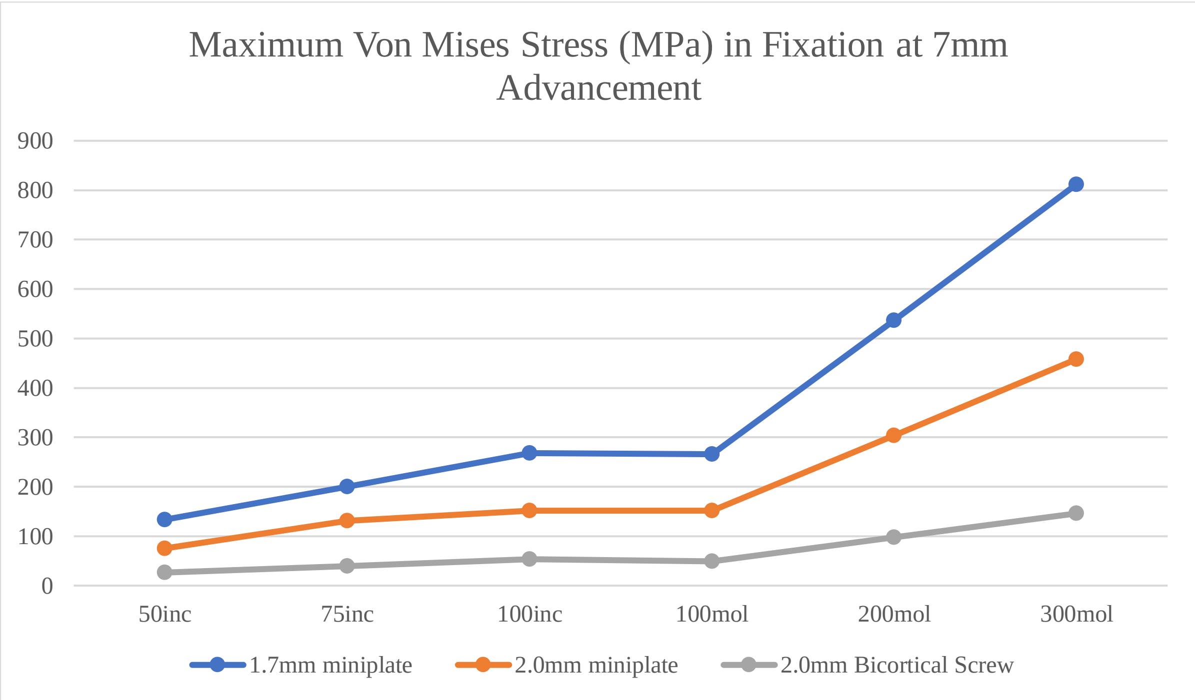


Figure 6.57 Stress in fixation techniques at 7mm Advancement

Shapiro-Wilks test was used to test normality of the data obtained from maximum stress in fixation techniques. Data is considered normal when p-value >0.05. The results for each fixation techniques are shown in the table below and illustrated with histogram diagrams. The results of the Shapiro-Wilk test and histogram illustration shows that the data for each fixation technique group does not follow a normal distribution.

Table 6.5 Data normality test for maximum stress in fixation techniques using Shapiro-Wilk test

Fixation Technique	Shapiro-Wilk Normality Test	Conclusion
1.7mm miniplate	W = 0.84532, p-value = 0.007151 ($p<0.05$)	Non - normally distributed data
2.0mm miniplate	W = 0.86164, p-value = 0.013 ($p<0.05$)	Non - normally distributed data
2.0mm bi-cortical screws	W = 0.88629, p-value = 0.03336 ($p<0.05$)	Non - normally distributed data

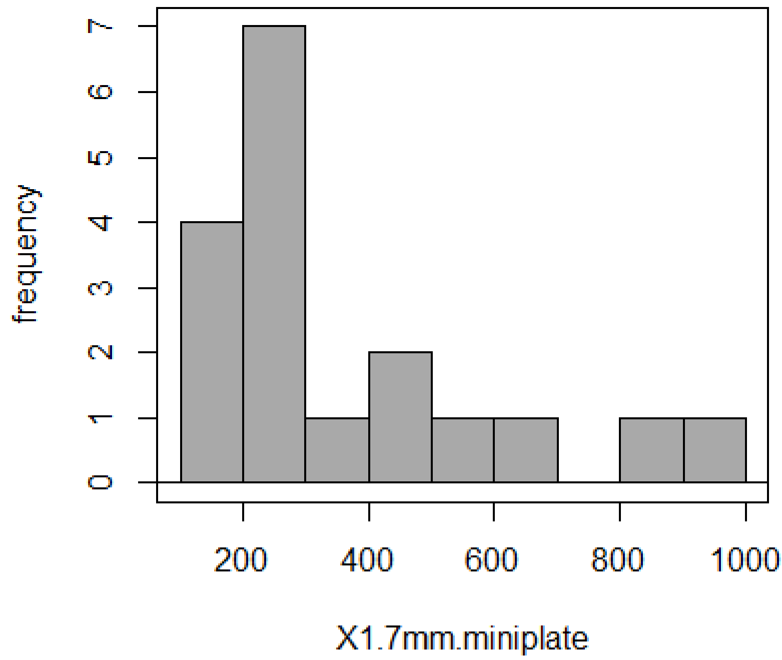


Figure 6.58 Data histogram for maximum stress in 1.7mm miniplate

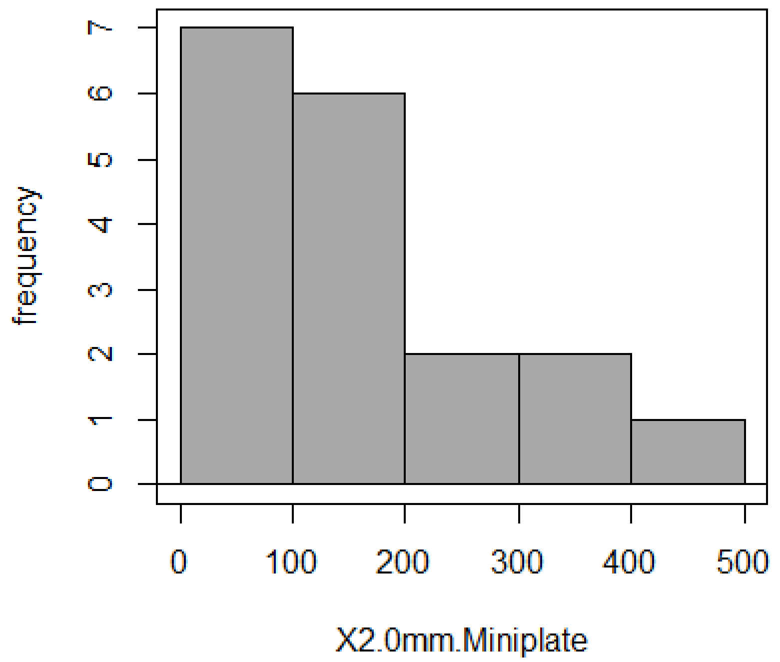


Figure 6.59 Data histogram for maximum stress in 2.0mm miniplate

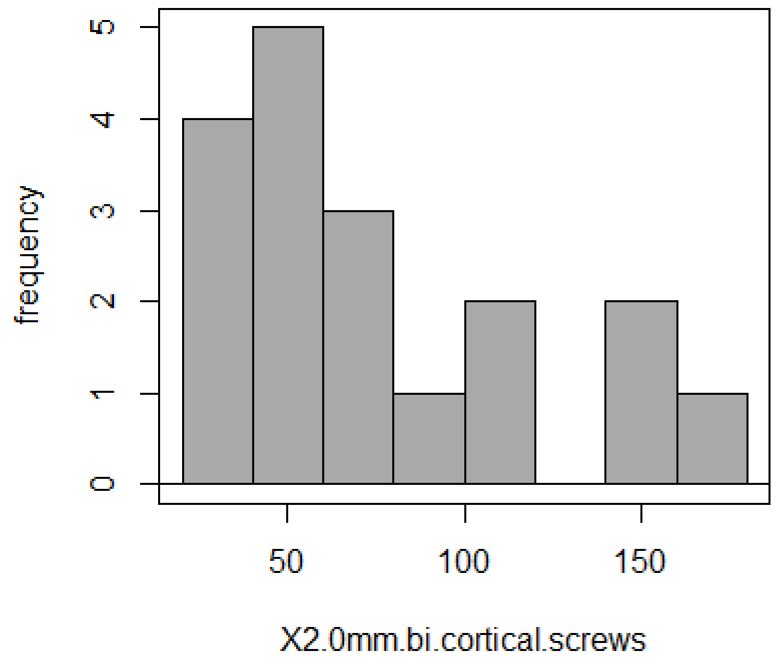


Figure 6.60 Data histogram for maximum stress in 2.0mm bi-cortical screws

Based on non-normal distribution of the results, a Wilcoxon rank sum test was carried out to compare the stresses between each fixation technique using the R Commander software. This is a non-parametric test, as the data does not follow a normal distribution, which is also an alternative to the two-sample t-test. The null hypotheses of the statistical test is when distribution of values in group 1 = the distribution of values in group 2. The null hypotheses is rejected when p-value < 0.05.

The test showed statistically significant difference between the 1.7mm miniplate and the 2.0mm miniplate, with $P < 0.05$ *(W = 263, p-value = 0.0009862)*. There was also significant difference between the 1.7mm and 2.0mm Bi-cortical screws, $P<0.05$ (*W = 314, p-value = 3.063e-08*), and between the 2.0mm miniplate and 2.0mm Bi-cortical screws, $P<0.05$ (*W = 238, p-value = 0.01557*).

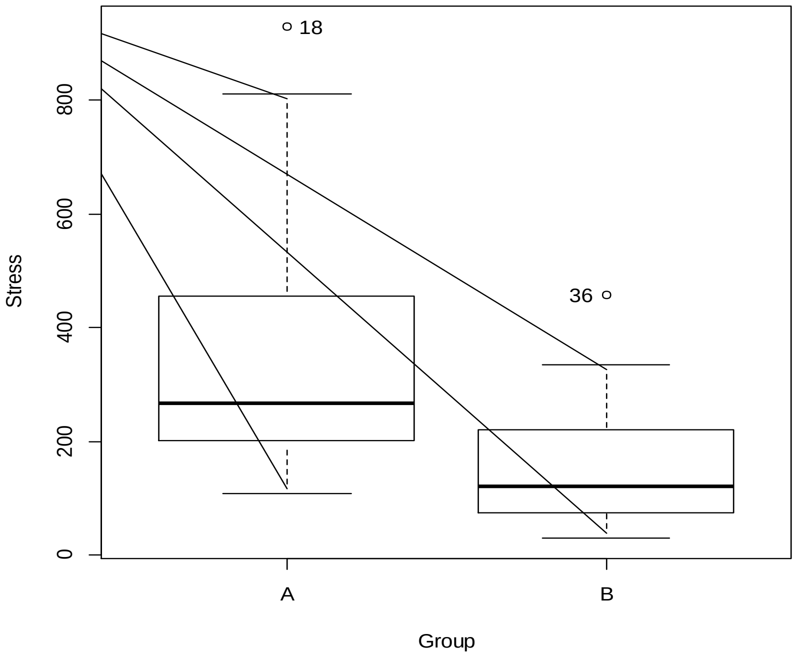


Figure 6.61 Box plot showing comparison of stress in Groups A (1.7mm miniplate) and B (2.0mm Miniplate)

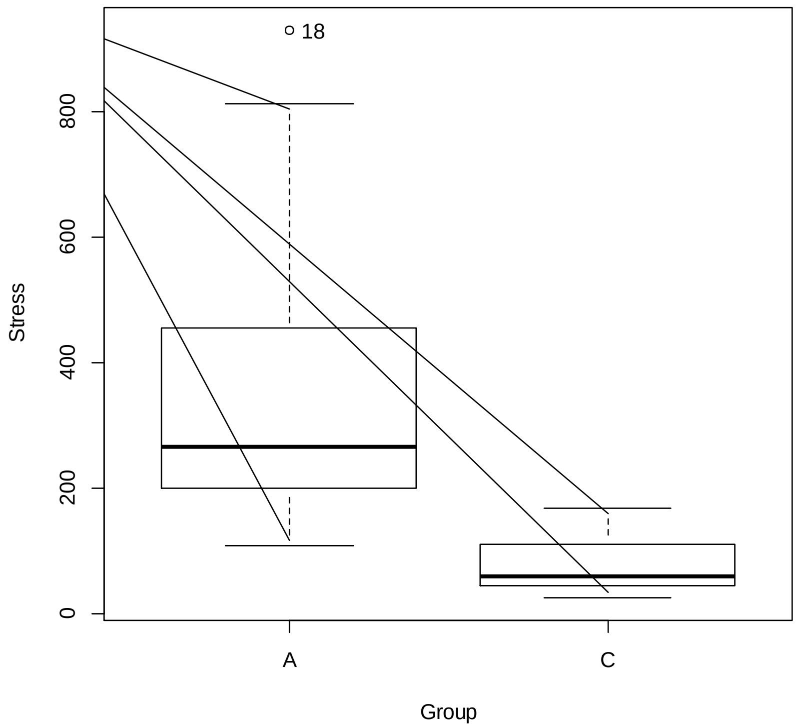


Figure 6.62 Box plot showing comparison of stress in Groups A (1.7mm miniplate) and C (2.0mm Bi-cortical Screws)

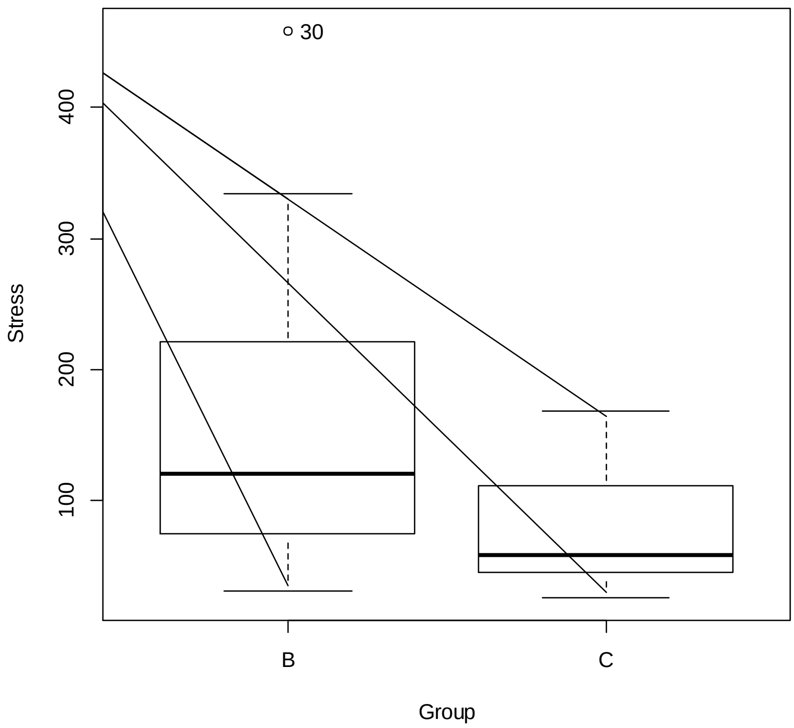


Figure 6.63 Box plot showing comparison of stress in Groups B (2.0mm miniplate) and C (2.0mm Bi-cortical Screws)

6.3.5 Comparison of stress in surrounding bone of fixation techniques

Stresses in the surrounding cortical bone were also recorded, and showed variable values in different fixation techniques, and in mandibular movements.

At 3mm setback, the 1.7 mm miniplate and 2.0mm miniplate recorded higher stresses in the cortical bone when compared to the 2.0mm bi-cortical screws.

As the mandible is advanced forward, the 2.0mm bi-cortical screws fixation were seen to have the highest stress in cortical bone, when compared to the other miniplate fixation techniques.

The maximum stress in cortical bone was recorded in Model 9 (2.0mm Bi-cortical Screws at 7mm mandibular advancement) at 300N force, which was seen in the surrounding distal cortical bone of the inferior – distal screw.

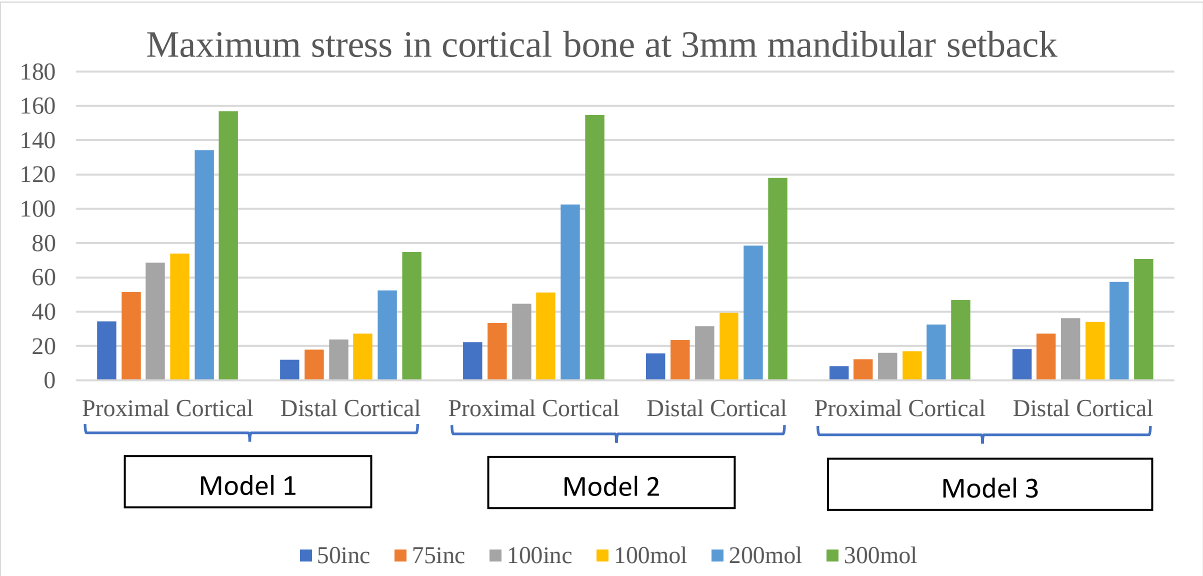


Figure 6.64 Stress in cortical bone surrounding fixation at 3mm mandibular setback

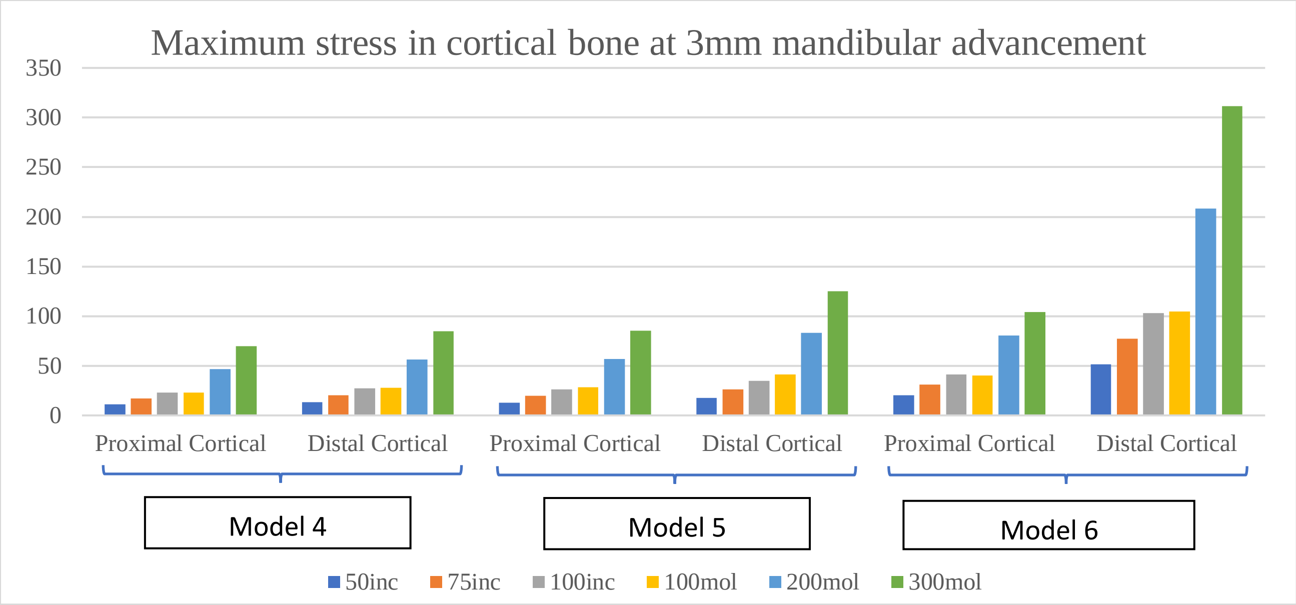


Figure 6.65 Stress in cortical bone surrounding fixation at 3mm mandibular advancement

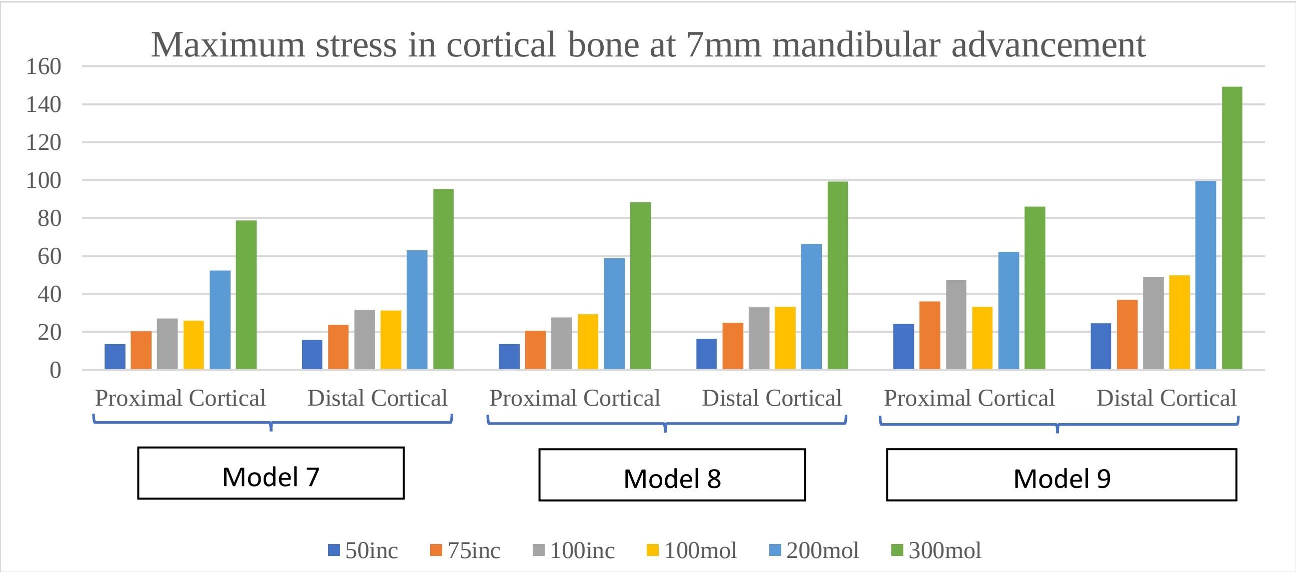


Figure 6.66 Stress in cortical bone surrounding fixation at 7mm mandibular advancement

6.3.6 Comparison of displacements in fixation techniques

In all models, the area of maximum displacement is seen in the angle region of the proximal mandible segment. This is also the region of force application. As applied force is increased, so does the resultant displacement increase. This is true for all mandibular movements and fixation techniques. The only exception is that the displacement in each model at 100N force and molar restraint is slightly lower than the 100N force, and with incisal restraint.

Models with 1.7mm miniplate have the highest displacement readings for all mandibular movements at each force magnitude, while the 2.0mm bi-cortical screws records the least displacement.

The maximum displacement for each for each fixation technique is increased when longer mandibular movement is applied.

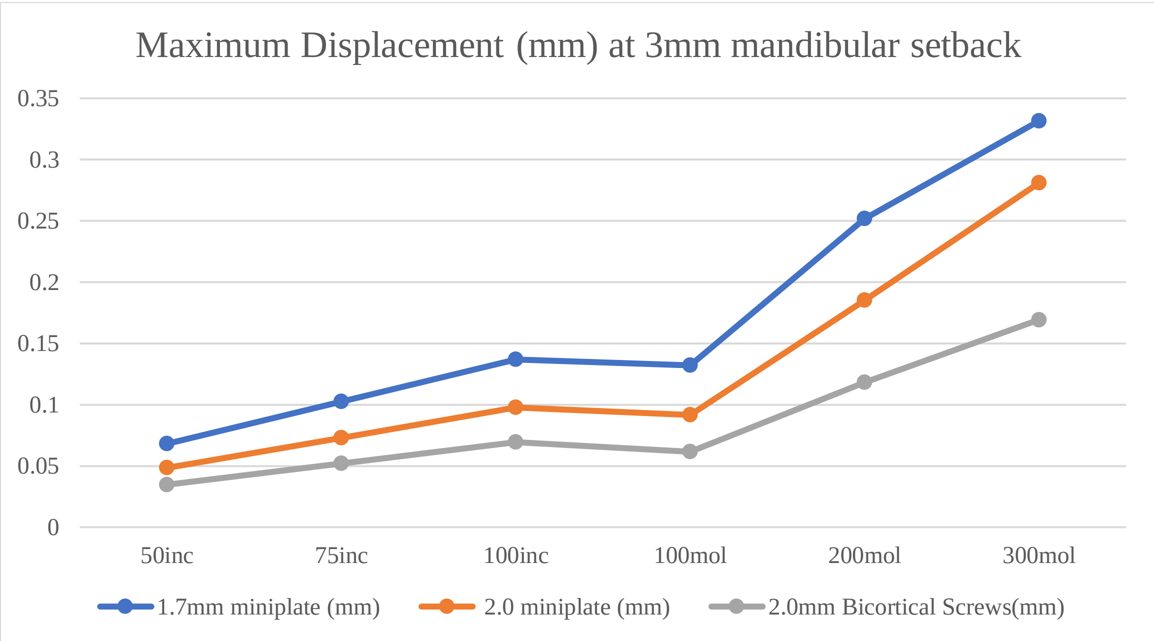


Figure 6.67 Mandibular displacement at 3mm mandibular setback

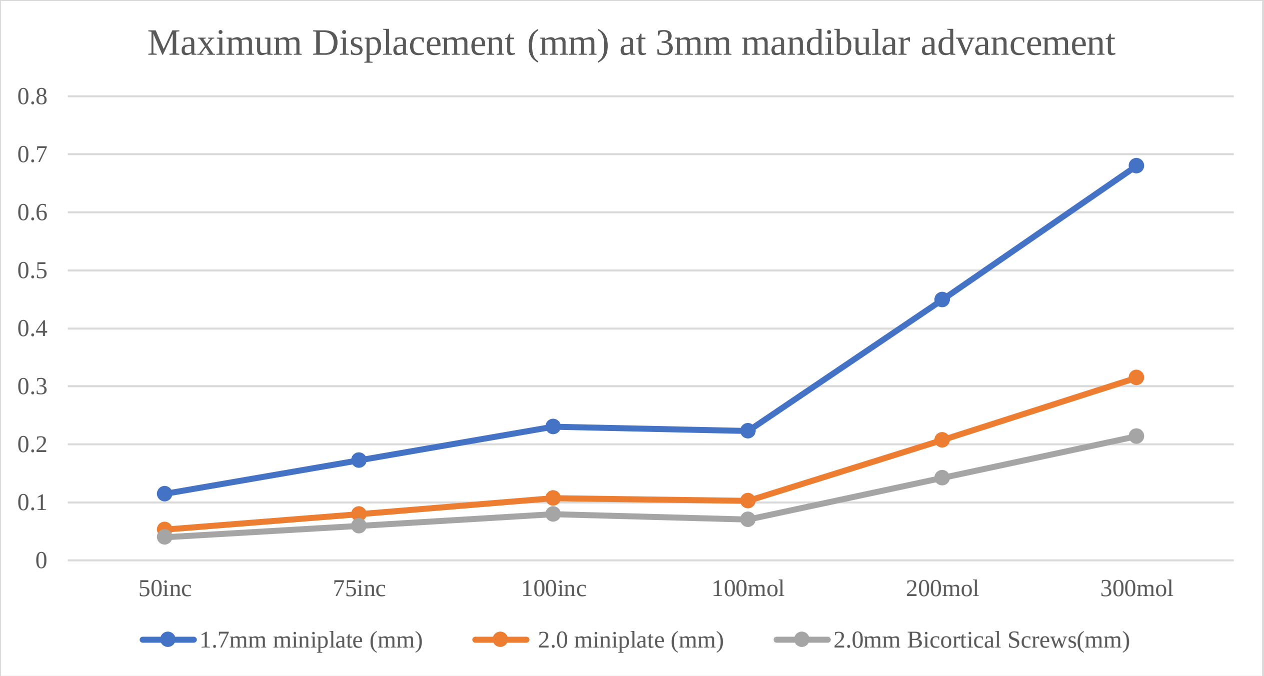


Figure 6.68 Mandibular displacement at 3mm mandibular advancement

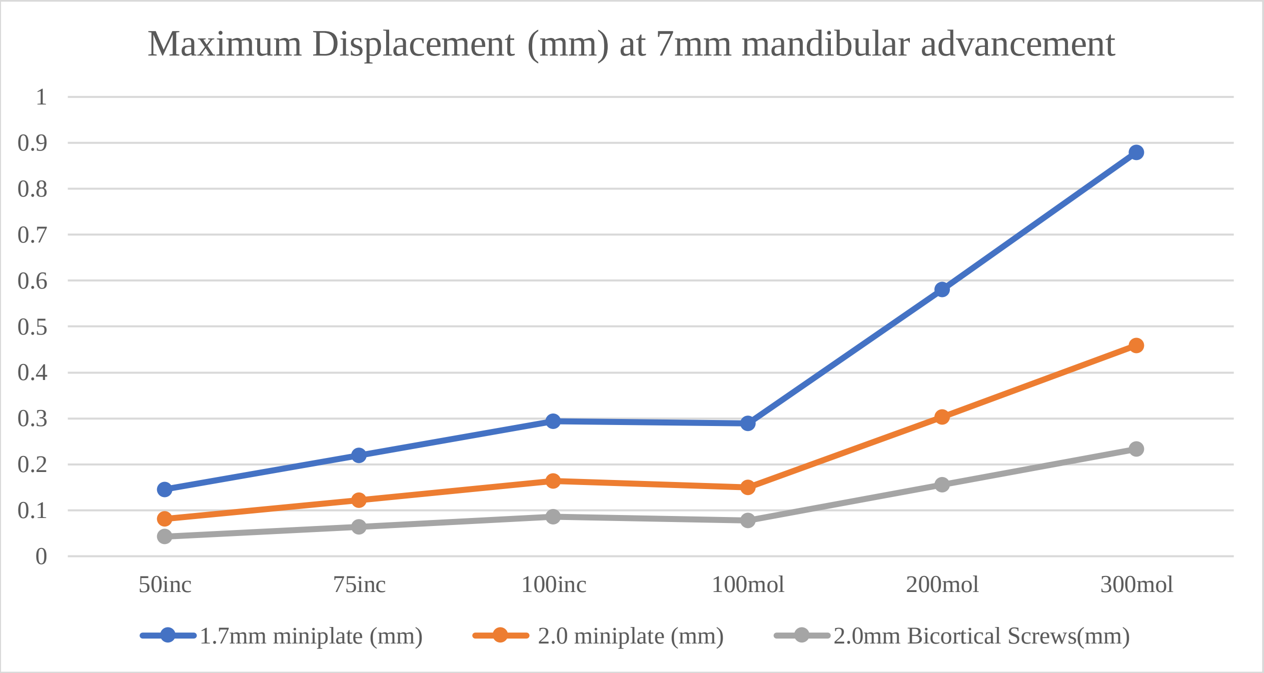


Figure 6.69 Mandibular displacement at 7mm mandibular advancement

Shapiro-Wilks test was used to test normality of the data obtained from maximum displacement in all simulation models. Data is considered normal when p-value >0.05. The results for each fixation technique groups are shown in the table below and illustrated with histogram diagrams. The results of the Shapiro-Wilk test and histogram illustration shows that the data for each fixation technique group does not follow a normal distribution.

Table 6.6 Data normality test for maximum displacement in all fixation technique groups using Shapiro-Wilk test

Fixation Technique Group	Shapiro-Wilk Normality Test	Conclusion
1.7mm miniplate	W = 0.83157, p-value = 0.004396 ($p<0.05$)	Non - normally distributed data
2.0mm miniplate	W = 0.85167, p-value = 0.008999 ($p<0.05$)	Non - normally distributed data
2.0mm bi-cortical screws	W = 0.85498, p-value = 0.01016 ($p<0.05$)	Non - normally distributed data

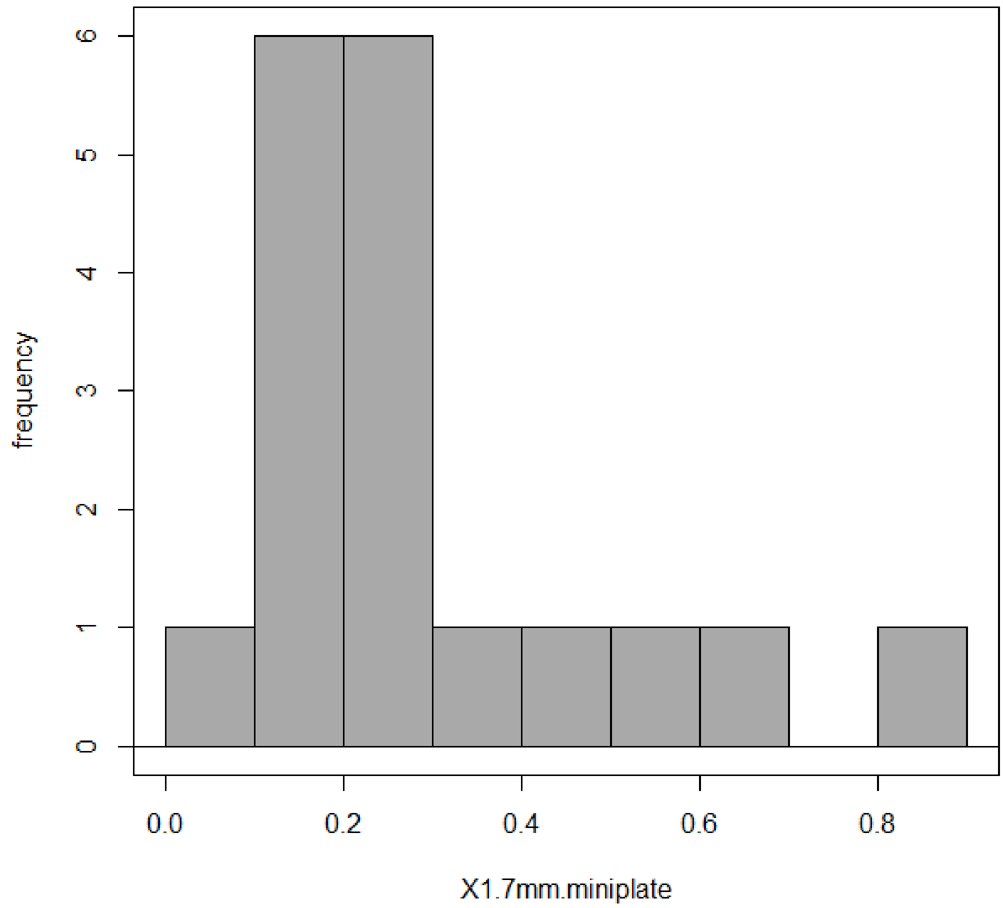


Figure 6.70 Data histogram for maximum displacement in 1.7mm miniplate fixation group

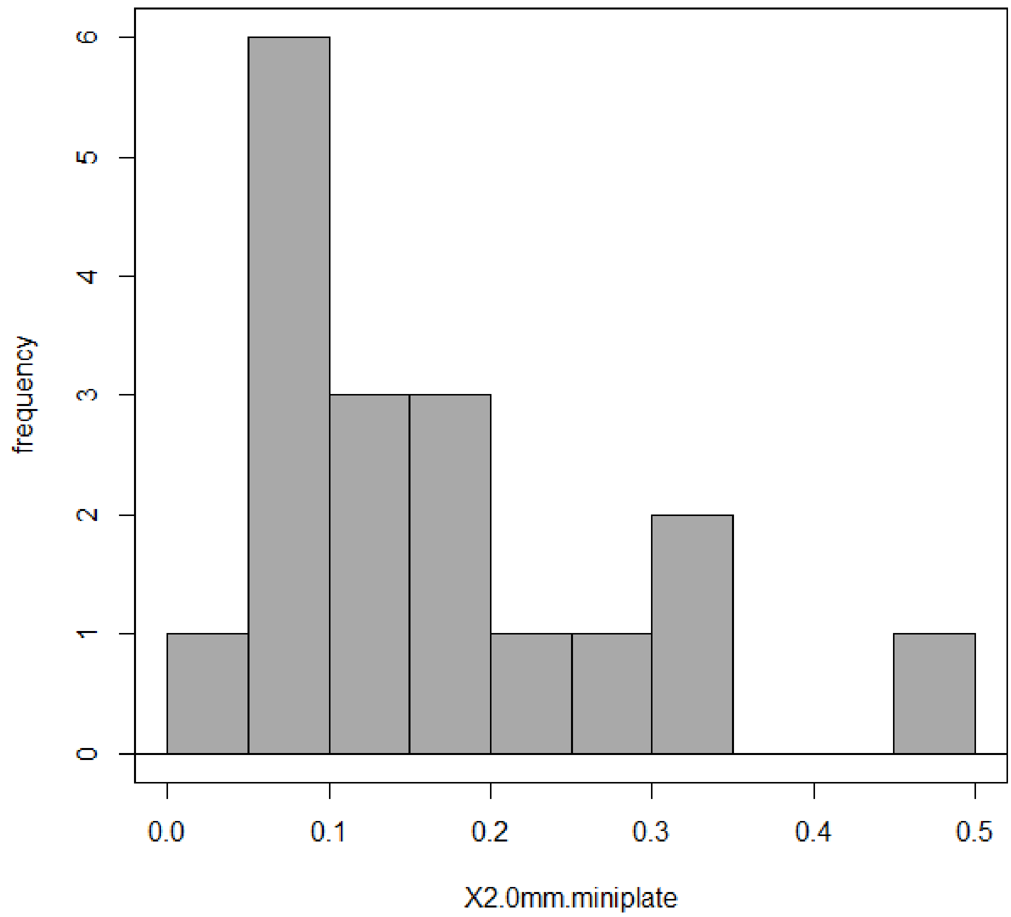


Figure 6.71 Data histogram for maximum displacement in 2.0mm miniplate fixation group

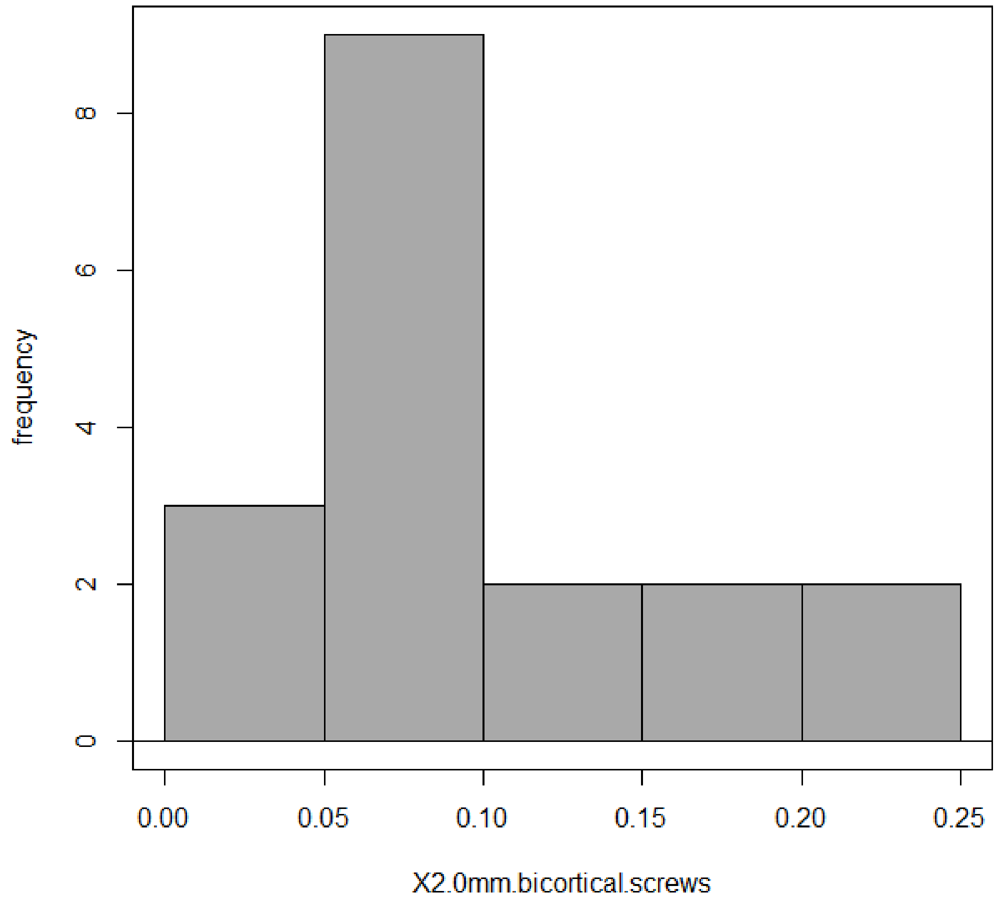


Figure 6.72 Data histogram for maximum displacement in 2.0mm bi-cortical screws fixation group

Based on the non-normality of the data distribution, a Wilcoxon rank sum test was carried out. Wilcoxon rank sum test have shown significant difference in displacement values between all fixation technique.

The test showed statistically significant difference between the 1.7mm miniplate and the 2.0mm miniplate, with $P < 0.05$ *(W = 235, p-value = 0.02046)*, between the 1.7mm and 2.0mm bi-cortical screws, $P<0.05$ (*W = 282, p-value = 5.811e-05*), and between the 2.0mm miniplate and 2.0mm bi-cortical screws, $P<0.05$ (W = 231, p-value = 0.02896). These are illustrated in the boxplots below.

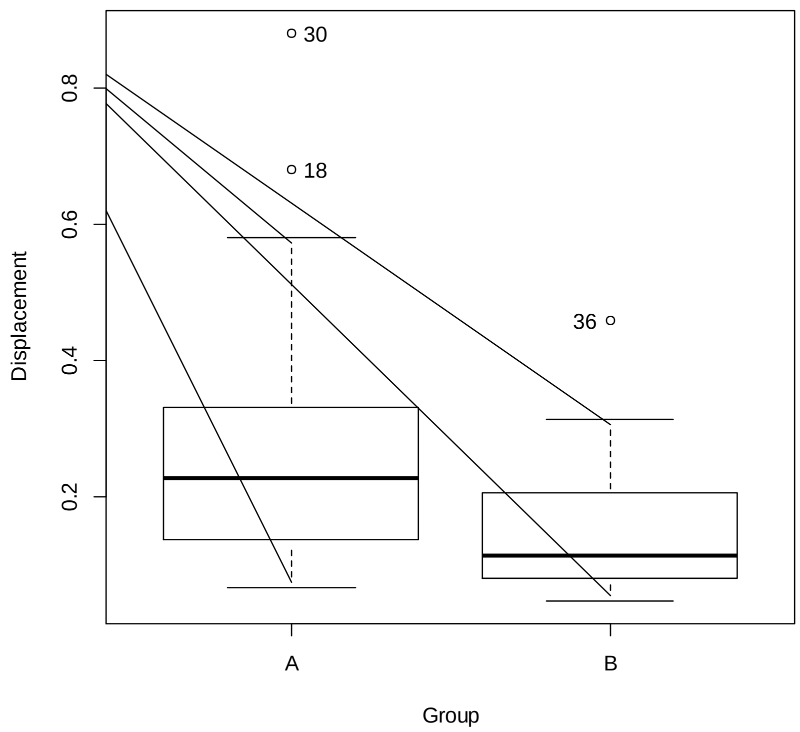


Figure 6.73 Box plot showing comparison of displacement in Groups A (1.7mm miniplate) and B (2.0mm miniplate)

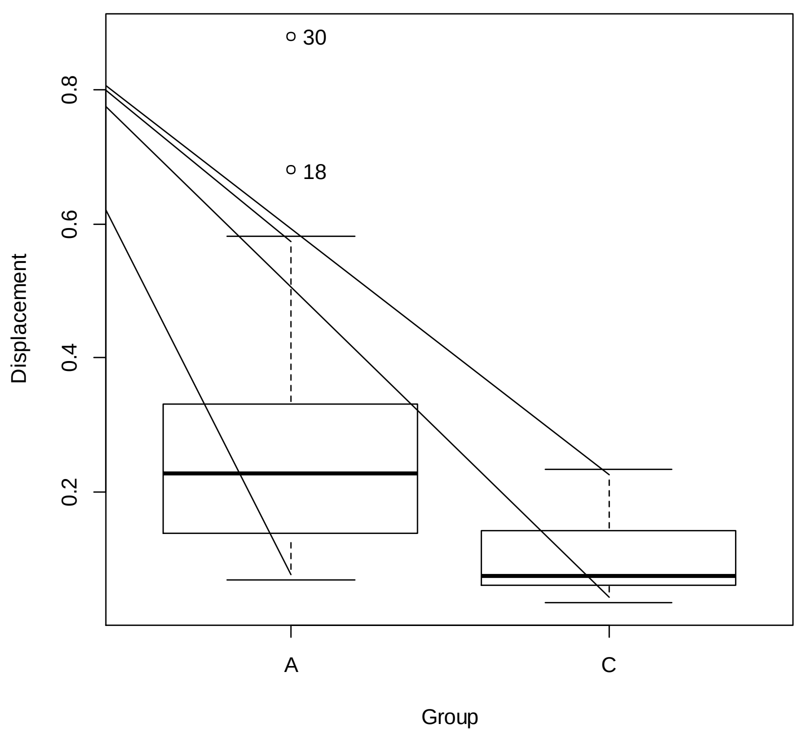


Figure 6.74 Box plot showing comparison of displacement in Groups A (1.7mm miniplate) and C (2.0mm Bi-cortical screws)

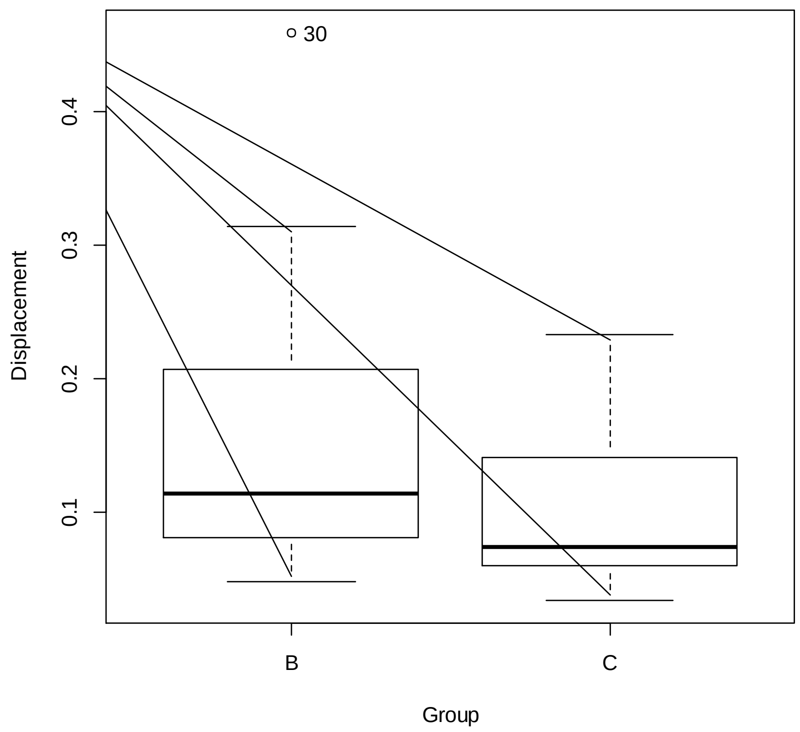


Figure 6.75 Box plot showing comparison of displacement in Groups B (2.0mm miniplate) and C (2.0mm Bi-cortical screws)

7. DISCUSSION

7.1 Model construction

The mandible model in this study was constructed in the equivalent manner as previous studies. A CT image of a mandible from a live patient was used to develop a 3-D virtual mandible model, which was used and manipulated in the study. The scanned mandible was only able to record the external surfaces perfectly, whereas the inner cortex and periodontal tissues could not be properly developed from the CT image. This did not pose an issue, as previous FEA studies had opted for a more simplified model in their studies. Maurer et al. in 1999 developed a hemi-mandible model, cortical and cancellous bone layers, and no teeth consisting of 19845 elements (39). Some studies considered isotropy in the mandible bone and did not include teeth in their models for simplicity(24, 37). In 2016, Stringhini et al. developed a full mandible model which included the cortical and cancellous bone, teeth, pulp, and periodontal tissues, which consisted of 1489170 elements (47). The model's complexity requires more computer resources, and longer running time to carry out the analyses. The relevance of including dental and periodontal tissues may be ideal, but questionable, when the analyses have no direct effect on these areas.

Therefore, in this study, the model created was a half mandible consisting of the full mandible dentition, and cortical and cancellous bone layers. Despite not having some anatomical features, such as the periodontal tissues, the model was constructed with a high number of elements which will be discussed further in the next chapter. The fixation techniques were developed in Solidworks using measurements provided from the manufacturer Stryker ©. The screws were developed as simple cylinders and screw threads omitted.

The choice of material properties was also important during model construction. Material properties were taken as average values from studies by Dechow et al 1993, Misch et al. 1999, Giesen et al 2001, and Scwartz-Dabney 2003 (54, 56-58). This is to ensure that the model developed can simulate reality. Dechow et al. explained that the variations of bone geometry and direction can affect the bone properties and elastic modulus. Bone in the longitudinal direction is stiffer, compared to the tangential and radial directions. While this is possible to develop in the FEA model, it does create more complexity. The material properties for all fixation techniques was constant. The type of titanium alloy used is Ti-6Al-4V, and its complete properties available in the Solidworks software.

This study applied bite forces from 50N to 300N. The amount of forces was chosen based on reports of bite forces following mandibular orthognathic surgery(59, 60, 72). Previous studies applied forces up to 600N in the posterior teeth region(21, 23). Although the authors wanted to reflect normal bite forces as reported in the literature, this amount of force application is unnecessary when comparing biomechanical properties of fixation techniques as it does not represent post-operative bite forces.

7.2 Simulation Models

In FEA, the way the models are simulated affects the results. This means that the location of force application, the magnitude of force, the restraints location and type of surface contact properties determine the results obtained from the analysis. FEA allows the user to restrain any surface on the model and apply force at any direction. However, this may be limited in in-vitro studies. In-vitro studies that compare biomechanical properties of fixation techniques uses a mandible model that is fixed onto a steel jig, usually in the condylar or ramus region. Force is applied using a hydraulic pressure guide onto the occlusal surface. The outcome is measured in displacement distance or fixation failure (deformation or fracture). Most FEA studies followed this method (20-24, 28, 37, 39, 40). However, this does not replicate the true physiological movement of the mandible. Bite force studies are carried out with participants biting onto a pressure sheet in maximum intercuspation (60). At this moment, forces are generated by the elevator muscles of mastication, namely the masseter, temporalis and medial pterygoid muscles(62). An FEA study by Oguz et al. in 2009 applied forces to the angle of mandible to simulate the masseter force while Stringhini et al. in 2016 applied forces from the masseter, medial pterygoid and temporalis (41, 47).

This study compared 6 different FEA simulations on Assembly Models 4, 5 and 6 (1.7mm miniplate, 2.0mm miniplate and 2.0mm bi-cortical screws at 3mm mandibular advancement). In all simulation models, the condyle was fixed in all surfaces, to fix the mandible in space while the mid-symphysis was restrained with the slider-roller restraint to simulate an identical mirrored side of the mandible. Due to this, condylar torque effect could not be studied and omitted from the study. The symphysis can only move in the vertical plane with the slider / roller restraint function.

The amount of stresses and displacement in all simulation models were observed. The highest stresses and displacement was recorded in Simulation Model B. This model was restrained in the condylar region only and had bite forces of 100N dispersed on the occlusal surface of the 1st molar, premolar and incisor at the ratio of 50:35:15 respectively. The stress recorded in each fixation was 1062N, 842N and 296.1N respectively. The displacement recorded in this model was 4.184mm, 3.432mm, and 2.37mm respectively. The lack of restraint in the model and increased surface of force application was the cause for high stress, and displacement readings, which may not represent clinically. When looked closely, stress areas of 30 - 60 MPa were seen in the anterior condylar neck region which indicates tension in the area (Fig 7.1)

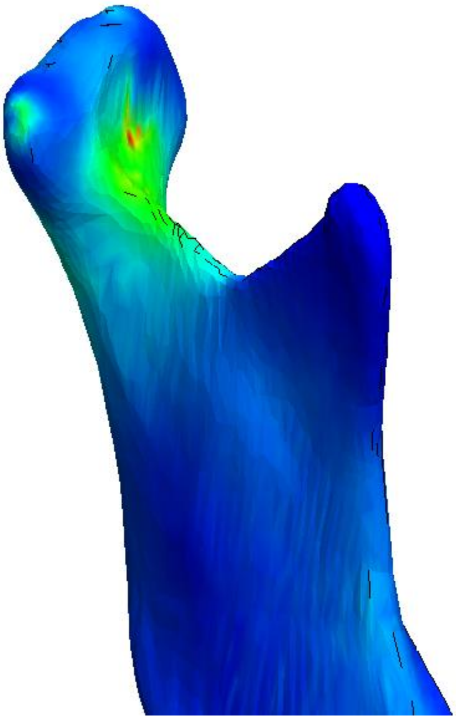


Figure 7.1 Stress in condylar neck region when force is loaded vertically onto teeth occlusal surface.

Simulation model's C and D recorded less stress and displacements as these models were similar to model's A and B but were restrained in the angle region. It is anticipated that these models will record higher displacement and stress values when more load is applied.

Simulation model's E and F recorded the least stress and displacement values. The force was applied in the angle region of the mandible, and either the molar or incisor was fixed. Restraining the respective tooth was to simulate biting conditions in respective areas. The displacement values were less than 1mm in both simulation model's E and F, which can be related to the direction of force application and restraint on both proximal and distal segments of the mandible. It was also seen that stresses in these models dissipate to other areas of the mandible (Fig 7.2). Stresses of 10 – 25 MPa were see in the condylar region and molar tooth where each were fixed. This implies that during biting, the forces generated by the muscle do not only transfer to the fixation hardware but also surrounding bone, teeth and more likely soft tissues. As this study's model does not include soft tissues, it is safe to assume that the actual stresses absorbed by the hard tissues are less in reality.

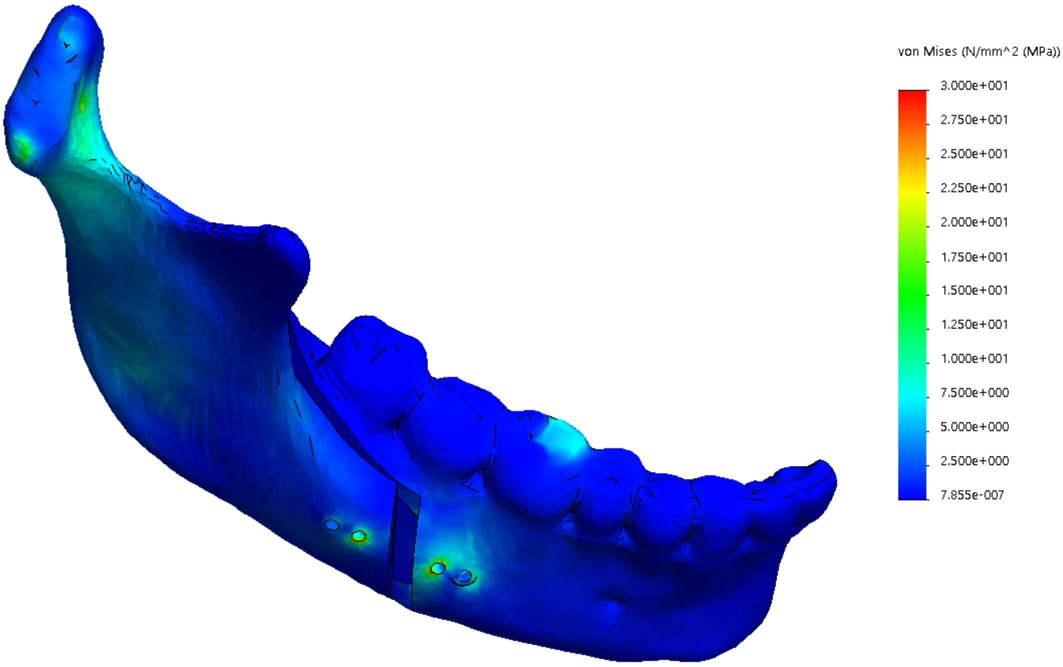


Figure 7.2 Stress areas in bone and teeth for Simulation Model E. Green to red colour indicate areas of stress – Condylar neck, occlusal surface, and surrounding fixation hardware.

The stress values also correlate with the amount of displacement in the model. This represents similar results by Stringhini et al. 2016 (47). While direct comparisons cannot be made between studies, the results were reassuring and encouraged our approach in using these simulation models further in the study. Simulation model E was used to study the fixation techniques at incisal bite forces of 50N, 75N and 100N while model F investigated molar bite forces of 100N, 200N and 300N (Fig 7.3).

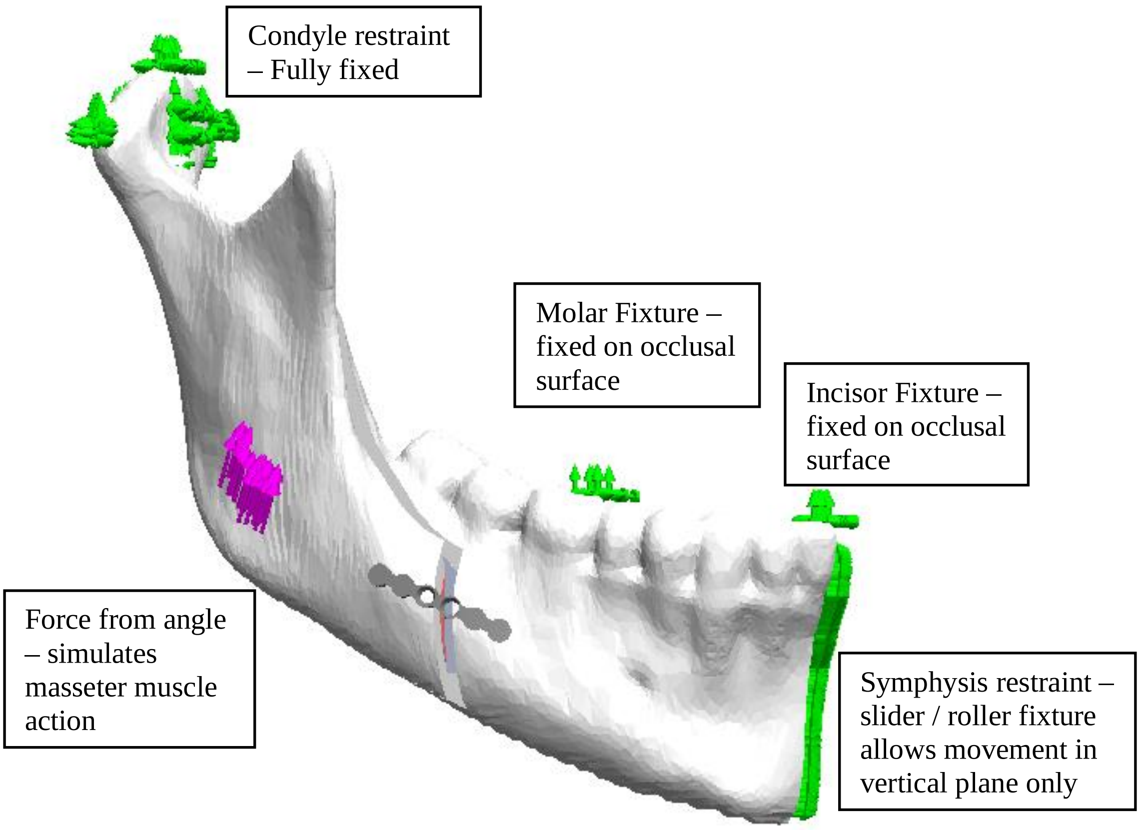


Figure 7.3 The chosen simulation model to run further simulations when comparing the three different fixation techniques. Either molar or incisor fixture is applied at one time with relevant force magnitudes.

7.3 Mesh Convergence Test

Anderson et al. 2007 outlines the importance of verifying and validating computational models in biomechanical studies (73). Verification of the model ensures that the model contains sufficient elements to approximate a continuum solution, which can be carried out with a mesh convergence test with a limit of maximum difference of 5%.

This study carried out a convergence test on Model 4 at 100N force (Fig 5.15, page 49). This model was chosen as it includes the smallest fixation technique which is the 1.7mm miniplate. Four mesh models underwent FEA simulation. The first model A consisted of a maximum and minimum element size of 6mm and 1mm respectively. The analysis on this model have shown that the area of maximum stress concentrates in the miniplate, which is 265.6 N. To validate this result, a mesh convergence test was required.

One way of validating is by refining the element size of the whole model until less than 5% difference between coarser and finer mesh is achieved. This method is time consuming, and requires more computer resources. Areas of less interest, for example interdental regions, do not require finer mesh, and areas of particular interest may not show accurate results. Alternatively, a local refinement command can be applied to certain parts of interest in the model. This allows manipulating the element sizes on chosen part(s) without disrupting the overall element size in the model. Therefore, by knowing the area of high stress is within the miniplate, a series of local refinement was carried out.

The results between model A without local refinement and models B, C, and D with local refinement was significantly different. Higher stresses were recorded when the local refinement was applied. Convergence was reached between models' C and D with less than 5% difference in results. This indicates that model C consists of the maximum element size required for the analysis to produce the best approximate result. Thus, further simulations in this study applied the mesh settings of model C.

7.4 Comparison of 3 fixation techniques

In this study, a comparison of 3 internal fixation techniques was carried out. The techniques include the 1.7mm miniplate, 2.0mm miniplate and 2.0mm bi-cortical screws. This study includes a smaller sized, multiple hole miniplate of 1.7mm, which is uncommon in mandibular fixation. The use of smaller miniplates in mandibular fixation has recently gained interest by clinicians and researchers. In 2010, Burm et al. studied the use of 1.2mm microplate and mono-cortical screws in the management of mandibular fractures and found considerable success compared to other conventional 2.0mm miniplates (74). He explained the advantages of smaller microplates, which include less foreign body reaction, less palpability, and improved patient acceptability. This study was supported by Ahmed et al. in 2017 who compared 2.0mm miniplates with 1.5mm miniplates in the management of mandibular symphysis fractures using FEA (75). He found that there was no significant difference in the stresses developed in either of the bone plating system. Yeo et al. compared the effects between 1.7mm upper border sliding plates and positional screws in neurosensory disturbances following BSSO advancement. He showed that there were insufficient evidence to prove one was better than the other (33).

7.4.1 Stress in fixations

Previous FEA studies have shown that bi-cortical screws exhibit less stress in the fixation when compared to miniplates(21, 24, 28, 37, 40, 45). This is also true in the present study. The 1.7mm miniplate have shown the highest stress in all mandibular movements compared to the 2.0mm miniplate and 2.0mm bi-cortical screws. The maximum stress recorded in the 1.7mm miniplate fixation in this study was within the ultimate strength of the material which is 950MPa (76). This means that the 1.7mm miniplate did not undergo material failure. The stress in the miniplate concentrates in the “connector region” or in the 1.7mm miniplate around the free hole regions. The 1.7mm miniplate is not manufactured with varying connector gaps such as the 2.0mm miniplate, as it is not intentionally produced for mandibular osteotomies. The peak stress is located at the superior - distal margin of the 3rd screw and inferior – mesial margin of the 2nd screw. This indicates high compression or tension stress in these areas. This similar stress pattern is also seen in the 2.0mm miniplate fixation.

The stress distribution pattern is different in the 2.0mm bi-cortical screws. The inferior distal screw shows more stress distribution, and peak stress is located in the lingual cortical – cancellous bone junction. Unlike the miniplates, the bi-cortical screws do not undergo as much bending. The reason of high stresses in the lingual region of the inferior distal screw is unclear but may be because it needs to withstand more force compared to the 2 screws superiorly.

From the results, miniplates are statistically significant less rigid than the bi-cortical screws, $p<0.05$ according to the Wilcoxon rank sum test. The penetration of both buccal and lingual cortices with the bi-cortical screws allows it to be more rigid, whereas the miniplates tend to undergo some degree of bending during force loading. This is true since the screws used in with the miniplates are mono-cortical.

The effects of various mandibular movements on fixation stress did not follow a certain pattern. Generally, the stress recorded at 3mm mandibular setback is less than the advancement movements. However, higher stress is recorded at the 3mm advancement compared to the 7mm advancement for the 1.7mm miniplate and 2.0mm bi-cortical screws. It was anticipated that as the mandibular advancement increases, the miniplate undergoes more bending and absorbs more stress. An explanation to this is that the FEA recorded the maximum stress in the element of the miniplate/screw rather than the average stress in the whole fixation. As for the 2.0mm miniplate, the stresses in this fixation increase with more mandibular advancement.

Model 1 (1.7mm miniplate at 3mm setback) at 300N force (fig 6.55) shows that the stress did not follow a linear pattern as compared to Models 2 and 3. Similar pattern is also seen with maximum displacement in Model 1 (fig 6.67). The 1.7mm miniplate have shown to allow more displacement compared to the 2.0mm miniplate and bi-cortical screws fixation. Thus, when increased force is applied in Model 1, the amount of bending that occurs in the 1.7mm miniplate causes approximation and contact between the proximal and distal mandible segments. Stresses are likely to be dispersed in this area of contact. Further displacement in the model is then affected by the boundaries set in the distal segment.

Findings in this study could not be compared to others as there is no available literature that compared effects of osteotomy distance on fixation stress. Erkmen et al. in 2005 published 2 studies comparing various fixation techniques on 5 mm SSO advancement and 5mm SSO setback (22, 23). However, the force applied was 500N and 660N respectively making the results invalid for comparison. The high force magnitude of 500 and 660N are also significantly over the normal bite force.

7.4.2 Stress in bone surrounding fixation

In this study, it is evident that the bone surrounding fixation helps absorbs stress, when force is loaded. The findings by Albougha et al. in 2015 (20) on stresses in bone surrounding fixation devices are similar to that of this study. Higher stresses and strains is seen in the bone with the bi-cortical screws when compared to the miniplate fixations. The exact location of stresses could not be compared because of the difference in FEA simulation and model construction. This study has shown that high stresses in bi-cortical screws is located around the inferior – distal screw. Maximum stress is seen in the lingual cortical bone around the screw (Fig. 7.4). This is also adjacent to the maximum stress seen in the screw fixation. This may also be related to the single screw present inferiorly compared to 2 screws superiorly. Stresses in bone surrounding the miniplate fixations are generally concentrated around the 3rd screw which is located at the buccal cortical surface of the distal mandible segment (Fig. 7.5 and 7.6).

However, this finding is only true in mandibular advancements. At 3mm mandibular setback, the miniplates recorded higher stresses compared to the bi-cortical screws. This is seen in the 2nd screw region of the proximal mandible segment. Generally, stresses in bone surrounding the fixation hardware are much higher than stresses in other areas of the mandible and teeth.

The relevance of bone stresses around the fixation hardware can be related to stress shielding. As metal is stiffer than bone, it becomes the main load supporter and reduces normal mechanical stress absorbed by bone. This can result in bone adapting to the new condition by reducing mass and resorption, and subsequently loosening of screws (77).

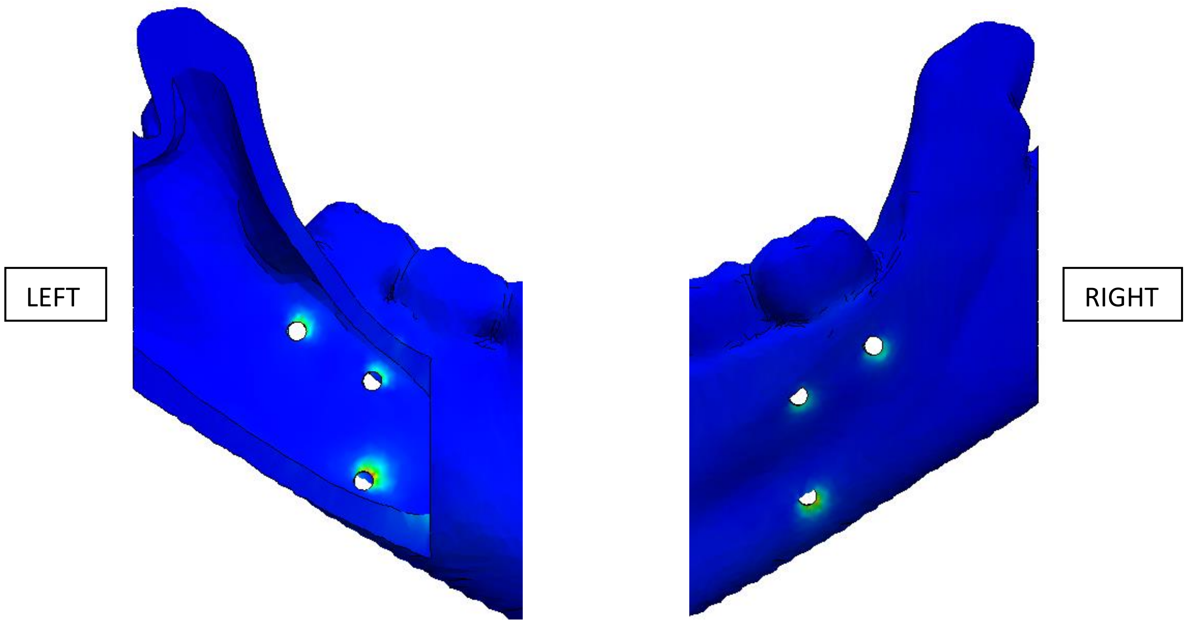


Figure 7.4 Stress in cortical bone surrounding bi-cortical screws. (Left) Buccal View of Distal Mandible Segment, (Right) Lingual View of Distal Mandible Segment

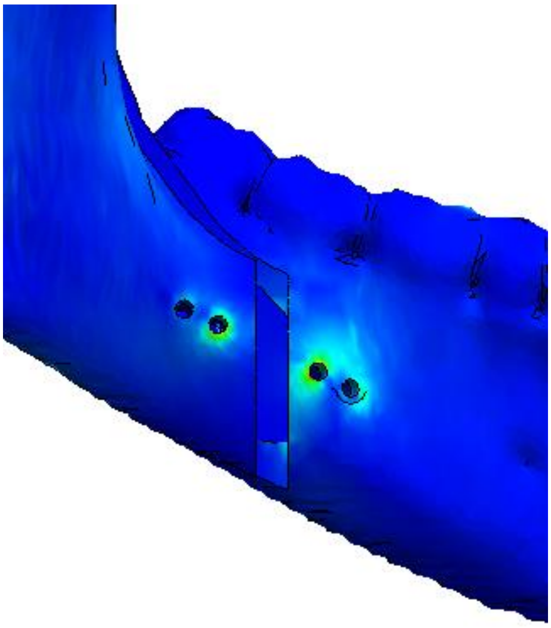


Figure 7.5 Stress in cortical bone surrounding 1.7mm Miniplate

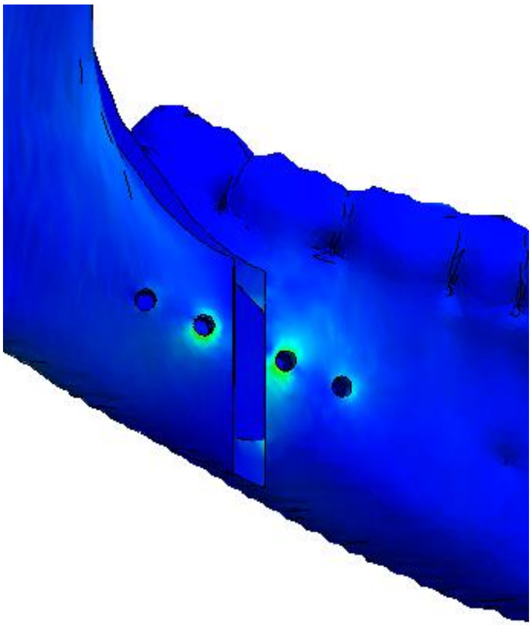


Figure 7.61 Stress in cortical bone surrounding 2.0mm Miniplate

7.4.3 Displacement of mandible bone segments

The amount of displacement of the mandibular segments indicates the rigidity and stability of the fixation technique. In the simulation models, the maximum displacement is located at the angle of the proximal mandible segment, where the force is applied (Fig 7.7). Displacement in this region is due to the muscle force pulling upwards, while the distal mandible segment is fixed with teeth at “maximum intercuspation”. There is also very little and insignificant movement of the chin.

Among all fixation techniques, the bi-cortical screws generate least displacement at all mandibular movements. There is a statistical significant difference between the bi-cortical screws, and miniplates fixation in the displacement of mandibular segments, with $p<0.05$ according to the Wilcoxon rank sum test. Previous in-vitro and FEA studies also report similar findings, where bi-cortical screws are more rigid than miniplate fixation(27, 29, 40). As the bi-cortical screws penetrates both cortices of the proximal and distal segment, it prevents rotational movement of the proximal segment. The 1.7mm miniplate and 2.0mm miniplate with mono-cortical screws allows some degree of bending and record more displacement readings when compared to the bi-cortical screws.

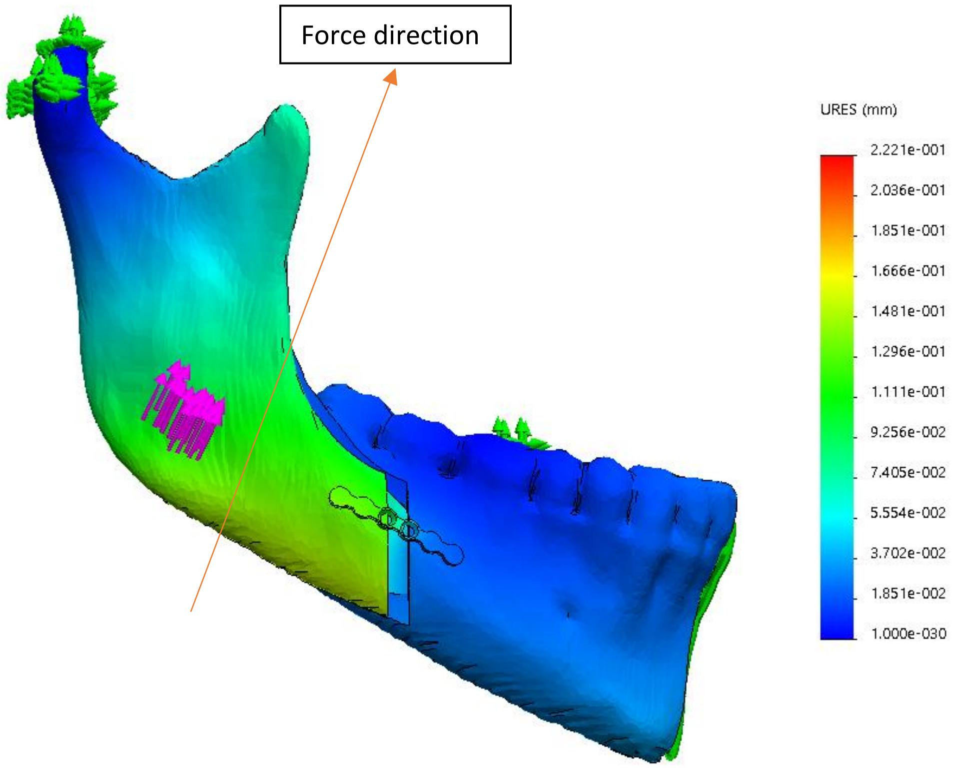


Figure 7.7 Displacement of mandible proximal segment when force is applied in direction of masseter muscle pull.

With increasing distances of mandibular advancement, the displacement increases for all fixation techniques. This is especially seen in the 1.7mm and 2.0mm miniplates. Less displacement is seen with 3mm mandibular setback as the proximal and distal bone segments are closely adapted and this increases stability, as explained in 7.4.1. Longer advances mean longer plates and distance between the proximal and distal segments increases, whereas the bi-cortical screws remain in the same position. Difference in displacement values were apparent in all fixation techniques, when the restraints on different teeth were applied. At 100N force, models with incisal restraints recorded higher displacement when compared with molar restraint. This may be related to the increased distance of the incisors to the force which permits more sagittal bending in the mandible.

Based on displacement values, the 1.7mm miniplate is the least rigid. This technique records a maximum displacement of 0.8797mm at 7mm mandibular advancement with 300N force. According to Claes in 1998, a rigid fixation minimizes interfragmentary movements and limited stimulation of callus formation, and a flexible fixation can enhance the callus formation, thus improving the healing process, whereas an unstable fixation can lead to a non-union (78). Furthermore, their animal experiments have shown that fracture gaps of >6mm will lead to non-union. This finding indicates an advantage for semi-rigid fixations, such as mini-plates in bone healing.

In the clinical setting, the choice of fixation techniques remains the surgeon’s preference. A systematic review by Al-Moraissi et al. 2016 have found that there were no statistically significant difference in skeletal stability between the bi-cortical screws and miniplate fixations (5). Although, the author did mention that the bi-cortical screws showed more skeletal stability, and would recommend the technique when advancing the mandible more than 6mm.

Joss et al. in 2009 carried out a systematic review investigating the stability of rigid internal fixation after bilateral sagittal split osteotomy advancement surgery (64). Interestingly, his findings show that the bi-cortical screws have more horizontal relapse compared to miniplates in the short and long term. However, relapse has multifactorial causes such as improper seating of condyles, amount of advancement or setback, mandibular plane angle, growth and others. Based on these systematic reviews, miniplates and bi-cortical screws have similar clinical success.

There has been no reported literature on the use of 1.7mm miniplates in mandibular osteotomies. Theoretically, the relevance of using smaller fixations includes less invasive surgery, reduce post-operative infection rates, damage to surrounding tissues and palpability. No clinical data is available to show the advantageous of smaller miniplates in mandibular osteotomy.

7.5 Limitations of FEA study

Limitations exist with any computational experiments. This study only involves the use of a single Caucasian female mandible CT scan. Although replicable and capable of being manipulated in the software, it may not represent a full population. Using data from more patients may strengthen the outcome.

The mandible model constructed was highly dependent on the accuracy of CT scan available. Due to this, structures such as the inferior dental canal could not be included. The inclusion of this structure would have influenced the placement of the fixation techniques and osteotomy cuts. The cancellous layer was also developed separately using the computer software, which might have caused anatomical inaccuracy.

Applying material properties to organic material as bone is difficult as there are various ranges between male and female, and bone elasticity changes with age. Furthermore, bone does not behave as an isotropic homogenous material. Efforts were made to ensure accurate material properties and best simulation method for the analysis.

In the model assemblies, the fixation screws were adapted closely to the bone and 'bonded' to ensure rigidity of screw placement. This is done as the screw threads were omitted in the design to reduce computational resources. However, fixation screws do not bond directly to bone in reality. Condylar torque was also not considered following fixation placement. It is a known phenomenon following fixation of the mandible. As the models did not include the temporomandibular joint structures and the condyle fixed during simulation, condylar torque is hard to appreciate.

This study only involved horizontal movements of the mandible, which was setback and advancement. This allowed the comparison of fixation techniques in various distances between the proximal and distal mandible segments. The distance of mandibular horizontal movement was measured in the Dal-Pont osteotomy (buccal vertical cut) between the proximal and distal mandible segments. Vertical and rotational movements also occur in BSSO but was not considered in this study as the objectives of the study was to compare the fixation techniques in mandibular setback and advancements only.

Although these limitations were present, they were mitigated by being present in all models that were compared with one another. The advantage of the FEA method is reproducing the same constant model throughout all simulations which help to eliminate construction error. In addition, this study attempted to replicate the physiological actions of the mandible at maximum intercuspation, and applied the finest mesh element size based on the convergence test. The results however, are comparisons between different internal fixation techniques and are not empirically correct.

8. CONCLUSION

With advancing technology and increasing availability in computer resources, the FEA method has very useful in biomechanical studies. Integration of computer engineering and the understanding of human physiology have given rise to a research tool that avoids the use of human or animal specimen, and inaccurate synthetic models. However, interpreting results from FEA studies must be taken with consideration.

This study investigated the biomechanical performance of three internal fixation techniques in the mandibular sagittal split osteotomy. A computer model of the mandible and fixation hardwares were developed with finite elements to compare these fixation techniques at variable mandibular movements and forces. A finite element mandible model and fixation devices were developed using computer software. The components were assembled to simulate various mandibular sagittal split osteotomy movements and fixed with different fixation techniques.

In this study, it was shown that most of the stresses were transferred to the fixations. The bi-cortical screws were more rigid compared to the 2.0mm miniplate and 1.7mm miniplate. Bi-cortical screws recorded the least stress and displacement. The 1.7mm miniplate was the least rigid fixation although the maximum stresses recorded were within its ultimate strength. The 2.0mm miniplate performed intermediately between the bi-cortical screws and 1.7mm miniplate. Generally, more stress and displacement are seen when the mandible is advanced compared to when it is set back. However, the difference between 3mm and 7mm advancement were not that significant.

The results have shown that smaller miniplates like the 1.7mm miniplate is able to provide sufficient fixation in various mandibular movements and within the reported post-operative bite forces. It is more stable in mandibular setback when the proximal and distal mandible is closely approximated. In advancements, the 1.7mm miniplate was able to withstand high forces and stresses absorbed by the hardware was within the material yield strength. Displacement values were also less than 1mm, making it more flexible than the 2.0mm miniplate and bi-cortical screws. This ‘flexibility’ can be an advantage post-operatively, where surgeons may improve the occlusion with intermaxillary fixation. It is safe to recommend the use of 1.7mm miniplate with mono-cortical screws in mandibular sagittal split osteotomy, although care must be taken when the advancement is >7mm, which was not investigated in the study. The restriction of soft diet is vital and application of intermaxillary fixation may be necessary.

More research and clinical data is required to compare the various type of fixation techniques on mandibular osteotomies, and their clinical outcomes. The results of this FEA should only be taken with consideration. In future, the choice of fixation will no longer depend on the surgeon’s surgical preference, but on further studies taking the intra-operative situation and patients’ best interest into account.

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APPENDIX

1. Simulation Data – 3mm mandibular setback

Force (N)	50	75	100	100	200	300
Tooth Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement (mm)	0.0681	0.1024	0.1369	0.1321	0.2516	0.3315
Max. Stress (MPa)						
1.7mm miniplate	107.90	162.00	216.30	229.80	406.30	456.30
Proximal Cortical	34.24	51.40	68.60	73.95	134.10	157.00
Distal Cortical	11.93	17.91	23.90	27.31	52.40	74.86
Proximal Cancellous	0.15	0.17	0.23	0.16	0.26	0.34
Distal Cancellous	4.37	6.56	8.75	13.74	26.32	35.54
Max. Strain (ESTRN)						
Fixation	1.47E-03	2.21E-03	2.94E-03	3.45E-03	6.40E-03	7.94E-03
Proximal Cortical	1.47E-03	2.20E-03	2.94E-03	3.35E-03	6.24E-03	8.10E-03
Distal Cortical	7.82E-04	1.64E-03	2.18E-03	1.76E-03	3.43E-03	4.88E-03
Proximal Cancellous	4.13E-04	6.17E-04	8.20E-04	5.81E-04	1.02E-03	1.49E-03
Distal Cancellous	5.29E-03	7.94E-03	1.06E-02	1.01E-02	1.81E-02	2.12E-02

Model 1 at 3mm mandibular setback

Force (N)	50	75	100	100	200	300
Tooth Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement (mm)	0.0485	0.0729	0.0975	0.0915	0.1851	0.2809
Max. Stress (MPa)						
2.0mm miniplate	30.67	46.06	61.51	75.80	152.20	229.20
Proximal Cortical	22.20	33.34	44.49	51.05	102.60	154.60
Distal Cortical	15.74	23.63	31.53	39.21	78.56	118.10
Proximal Cancellous	0.07	0.10	0.14	0.14	0.29	0.43
Distal Cancellous	0.44	0.66	0.87	0.95	1.90	2.86
Max. Strain (ESTRN)						
Fixation	3.94E-04	5.92E-04	7.90E-04	9.05E-04	1.81E-03	2.72E-03
Proximal Cortical	1.16E-03	1.74E-03	2.32E-03	2.63E-03	5.28E-03	7.93E-03
Distal Cortical	9.19E-04	1.37E-03	1.84E-03	2.30E-03	4.61E-03	6.95E-03
Proximal Cancellous	2.72E-04	4.09E-04	5.45E-04	4.35E-04	8.78E-04	1.32E-03
Distal Cancellous	1.53E-03	2.31E-03	3.09E-03	3.91E-03	7.86E-03	1.18E-02

Model 2 at 3mm mandibular setback

Force (N)	50	75	100	100	200	300
Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement(mm)	0.0347	0.0521	0.0695	0.0618	0.1182	0.1693
Stress (MPa)						
Fixation						
1. superior-proximal	14.34	21.49	28.52	34.04	61.21	84.88
2. superior-distal	13.84	20.75	27.52	30.78	54.74	71.81
3. inferior-distal	38.18	57.25	75.99	67.81	111.4	145.20
Proximal Cortical	8.122	12.18	16.17	16.94	32.47	46.90
Distal Cortical	18.22	27.31	36.24	34.21	57.27	70.70
Proximal Cancellous	0.21	0.31	0.41	0.45	0.80	1.04
Distal Cancellous	0.31	0.47	0.62	0.68	1.23	1.52
Strain (ESTRN)						
Fixation						
1. superior-proximal	1.34E-04	2.00E-04	2.66E-04	3.34E-04	6.14E-04	8.38E-03
2. superior-distal	1.31E-04	1.96E-04	2.61E-04	3.27E-04	5.86E-04	7.81E-04
3. inferior-distal	3.50E-04	5.24E-04	7.00E-04	7.39E-04	1.32E-03	1.75E-03
Proximal Cortical	6.46E-04	9.69E-04	1.29E-03	1.63E-03	2.90E-03	3.84E-03
Distal Cortical	1.26E-03	1.88E-03	2.50E-03	2.43E-03	4.20E-03	5.36E-03
Proximal Cancellous	3.01E-04	1.20E-03	1.60E-03	1.72E-03	2.99E-03	4.03E-03
Distal Cancellous	1.23E-03	1.84E-03	2.44E-03	0.00241	4.12E-03	5.25E-03

Model 3 at 3mm mandibular setback

2. Simulation data – 3mm mandibular advancement

Force (N)	50	75	100	100	200	300
Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement (mm)	0.1147	0.1723	0.2304	0.2234	0.4493	0.6800
Stress (MPa)						
1.7mm miniplate	148.70	223.50	298.80	307.50	621.30	929.70
Proximal Cortical	11.30	17.16	22.87	23.16	46.39	69.63
Distal Cortical	13.34	20.38	27.20	27.99	56.29	84.80
Proximal Cancellous	0.41	0.61	0.81	0.73	1.41	2.06
Distal Cancellous	3.40	5.15	6.93	6.72	13.91	21.66
Strain (ESTRN)						
Fixation	1.64E-03	2.48E-03	3.31E-03	3.34E-03	6.74E-03	1.08E-02
Proximal Cortical	9.60E-04	1.45E-03	1.94E-03	1.67E-03	3.41E-03	5.16E-03
Distal Cortical	1.18E-03	1.78E-03	2.37E-03	2.32E-03	4.64E-03	6.94E-03
Proximal Cancellous	1.32E-03	1.94E-03	2.58E-03	2.36E-03	4.57E-03	6.69E-03
Distal Cancellous	1.40E-02	2.12E-02	2.85E-02	2.75E-02	5.64E-02	8.68E-02

Model 4 at 3mm mandibular advancement

Force (N)	50	75	100	100	200	300
Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement (mm)	0.0530	0.0799	0.1069	0.1023	0.2071	0.3147
Stress (MPa)						
2.0mm miniplate	50.45	75.81	101.3	110.1	221.5	334.30
Proximal Cortical	13.01	19.54	26.08	28.28	56.79	85.55
Distal Cortical	17.48	26.24	35.02	41.39	82.99	124.80
Proximal Cancellous	0.08	0.13	0.17	0.167	0.34	0.51
Distal Cancellous	1.10	1.65	2.20	2.28	4.57	6.89
Strain (ESTRN)						
Fixation	5.06E-04	7.60E-04	1.01E-03	1.13E-03	2.27E-03	3.43E-03
Proximal Cortical	7.46E-04	1.12E-03	1.49E-03	1.59E-03	3.20E-03	4.81E-03
Distal Cortical	9.68E-04	1.45E-03	1.94E-03	2.28E-03	4.59E-03	6.92E-03
Proximal Cancellous	2.81E-04	4.24E-04	5.70E-04	5.48E-04	1.10E03	1.67E-03
Distal Cancellous	4.80E-03	7.24E-03	9.70E-03	9.67E-03	1.97E-02	3.02E-02

Model 5 at 3mm mandibular advancement

Force (N)	50	75	100	100	200	300
Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement(mm)	0.03983	0.05983	0.07989	0.07056	0.1418	0.2139
Stress (MPa)						
Fixation						
1. superior-proximal	12.65	18.98	25.30	28.70	57.35	85.96
2. superior-distal	12.02	18.03	24.02	29.96	59.82	89.58
3. inferior-distal	30.35	45.51	60.68	56.26	112.40	168.50
Proximal Cortical	20.54	30.87	41.23	40.15	80.73	104.30
Distal Cortical	51.70	77.49	103.20	104.50	208.30	311.30
Proximal Cancellous	0.21	0.31	0.42	0.48	0.96	1.44
Distal Cancellous	22.32	33.47	44.61	39.03	77.95	116.70
Strain (ESTRN)						
Fixation						
1. superior-proximal	1.56E-04	2.33E-04	3.11E-04	3.85E-04	7.68E-04	1.15E-03
2. superior-distal	1.40E-04	2.09E-04	2.79E-04	3.05E-04	6.09E-04	9.14E-04
3. inferior-distal	2.01E-03	3.02E-03	4.03E-03	3.69E-03	7.39E-03	1.11E-02
Proximal Cortical	1.21E-03	1.82E-03	2.43E-03	1.71E-03	3.41E-03	5.13E-03
Distal Cortical	4.73E-03	7.14E-03	9.52E-03	9.69E-03	1.94E-02	2.93E-02
Proximal Cancellous	7.31E-04	1.10E-03	1.46E-03	1.70E-03	3.42E-03	5.13E-03
Distal Cancellous	1.57E-03	2.35E-03	3.14E-03	0.00363	7.24E-03	1.09E-02

Model 6 at 3mm mandibular advancement

3. Simulation data – 7mm mandibular advancement

Force (N)	50	75	100	100	200	300
Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement (mm)	0.1458	0.2194	0.2936	0.2887	0.5808	0.8797
Stress (MPa)						
1.7mm miniplate	133.50	200.6	267.80	266.30	537	812
Proximal Cortical	13.59	20.38	27.16	26.08	52.30	78.67
Distal Cortical	15.83	23.77	31.71	31.31	63.08	95.33
Proximal Cancellous	0.54	0.72	1.07	1.09	2.15	3.21
Distal Cancellous	2.87	4.33	5.79	5.70	11.58	17.65
Strain (ESTRN)						
Fixation	1.52E-03	2.29E-03	3.06E-03	2.94E-03	5.98E-03	9.28E-03
Proximal Cortical	8.41E-04	1.26E-03	1.21E-03	1.48E-03	2.75E-03	4.15E-03
Distal Cortical	9.12E-04	1.37E-03	1.83E-03	1.82E-03	3.66E-03	5.53E-03
Proximal Cancellous	1.83E-03	2.75E-03	3.67E-03	3.70E-03	7.20E-03	1.06E-02
Distal Cancellous	1.08E-02	1.62E-02	2.16E-02	2.13E-02	4.31E-02	6.55E-02

Model 7 at 7mm mandibular advancement

Force (N)	50	75	100	100	200	300
Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement (mm)	0.08132	0.1223	0.1636	0.1501	0.303	0.4593
Stress (MPa)						
2.0mm miniplate	75.31	131.10	151.10	151.20	304.10	458.10
Proximal Cortical	13.76	20.66	27.57	29.25	58.73	88.40
Distal Cortical	16.50	24.75	32.99	33.18	66.28	99.29
Proximal Cancellous	0.17	0.26	0.35	0.26	0.52	0.73
Distal Cancellous	1.12	1.68	2.25	2.11	4.26	6.44
Strain (ESTRN)						
Fixation	7.70E-04	1.15E-03	1.55E-03	1.58E-03	3.19E-03	4.82E-03
Proximal Cortical	7.45E-04	1.11E-03	1.47E-03	1.53E-03	3.14E-03	4.62E-03
Distal Cortical	7.73E-04	1.16E-03	1.55E-03	1.68E-03	3.38E-03	5.12E-03
Proximal Cancellous	1.13E-03	1.69E-03	2.26E-03	1.72E-03	3.39E-03	4.75E-03
Distal Cancellous	4.09E-03	6.15E-03	8.21E-03	7.48E-03	1.50E-02	2.26E-02

Model 8 at 7mm mandibular advancement

Force (N)	50	75	100	100	200	300
Restraint	incisor	incisor	incisor	molar	molar	molar
Displacement(mm)	0.0427	0.06411	0.08557	0.07738	0.1551	0.2333
Stress (MPa)						
Fixation						
1. superior-proximal	12.91	19.36	25.81	26.86	53.63	80.30
2. superior-distal	12.91	19.40	25.82	27.67	55.38	83.13
3. inferior-distal	26.45	39.68	52.91	48.66	97.32	146
Proximal Cortical	24.17	36.01	47.34	33.35	62.06	85.99
Distal Cortical	24.54	36.81	49.07	49.76	99.45	149.1
Proximal Cancellous	0.19	0.28	0.38	0.37	0.75	1.14
Distal Cancellous	0.261	0.39	0.52	0.53	1.06	1.59
Strain (ESTRN)						
Fixation						
1. superior-proximal	1.26E-04	1.89E-04	2.52E-04	2.85E-04	5.69E-04	8.51E-04
2. superior-distal	1.26E-04	1.95E-04	2.60E-04	2.77E-04	5.53E-04	8.28E-04
3. inferior-distal	3.39E-04	5.08E-04	6.78E-04	6.21E-04	1.24E-03	1.87E-03
Proximal Cortical	1.24E-03	1.84E-03	2.42E-03	1.70E-03	3.18E-03	4.41E-03
Distal Cortical	1.38E-03	2.08E-03	2.77E-03	2.80E-03	5.62E-03	8.46E-03
Proximal Cancellous	6.69E-04	1.00E-03	1.34E-03	1.52E-03	3.04E-03	4.58E-03
Distal Cancellous	8.36E-04	1.25E-03	1.67E-03	0.00163	3.53E-03	4.90E-03

Model 9 at 7mm mandibular advancement