FACTORS INFLUENCING FIXATION OF PLATES IN FRACTURE MANDIBLE – A CLINICAL & BIOMECHANICAL STUDY

Dissertation submitted to

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In partial fulfillment for the Degree of

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BRANCH III

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CERTIFICATE

This is to certify that this dissertation titled "**FACTORS INFLUENCING FIXATION OF PLATES IN FRACTURE MANDIBLE – A CLINICAL & BIOMECHANICAL STUDY"** is a bonafide record of work done by **Dr.V.KIRUTHIKA** under my guidance during her postgraduate study period 2013 - 2016.

This dissertation is submitted to **THE TAMILNADU Dr. M.G.R. MEDICAL UNIVERSITY,** in partial fulfillment for the degree of **MASTER OF DENTAL SURGERY** in **Branch III – ORAL AND MAXILLOFACIAL SURGERY**.

It has not been submitted (partially or fully) for the award of any other degree or diploma.

Professor, HOD and Guide

Dr. L.DEEPANANDAN, M.D.S.,
Department of Oral & Maxillofacial surgery,
Sri Ramakrishna Dental College & Hospital,
Coimbatore.

Principal

Dr. V. PRABHAKAR, M.D.S.,
Sri Ramakrishna Dental College & Hospital,
Coimbatore.

Candidate

Dr. V. KIRUTHIKA

Department of Oral & Maxillofacial surgery,
Sri Ramakrishna Dental College & Hospital,
Coimbatore.

Date:

Place: Coimbatore

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Dr. V. KIRUTHIKA

AIM: The symphysis and angle region are the most frequent sites for mandibular fractures. Direct application of 2.0mm conventional and locking titanium miniplates are the most commonly used intraoral open reduction and internal fixation technique today. Anatomic and biomechanical limitations continue to make this application technically challenging with a considerable complication rate. Such incongruences are analysed with respect to the complex biomechanical behaviour of the mandible.

METHODOLOGY: Individual human mandible geometry, the specific bone density distribution, and the position and orientation of the masticatory muscles were evaluated by performing computed tomography scan of the cadaveric human mandible. Dimensional changes in the holes of the 2.0mm (Orthomax, Leforte and Synthes) titanium conventional and locking miniplates/screws were evaluated using RAPID-I Precision Vision Measuring System (VMS) pre and post adaptation to angle and symphysis region. The average bite forces of 15 patients who were operated for symphysis and angle fractures were measured using AXPERT electronic bite force gauge at 5 bite points viz right molars, right premolars, left molars, left premolars and anteriors. Three Dimensional Finite Element Analysis (3D FEA) was performed for symphysis and angle fracture sites with Temporomandibular Joint remaining static. Deflection, stability, mechanical stress over bone, maximal stress over miniplate, fracture gap and direction of displacement evaluated for loading conditions.

RESULTS: Symphysis fracture fixation showed maximum deflection of 6.05196mm with Orthomax conventional and least of 2.50747mm with Leforte locking miniplates. Maximum stress over bone was 98.6587 Mpa with Orthomax conventional and least was with Synthes locking of about 78.476 MPa. Stress over plate was more of about 75.4011 MPa in Orthomax conventional and least of about 61.2447 MPa in Synthes locking. Fracture gap was more of about 0.86241mm in Orthomax conventional and least of about 0.01804mm with Leforte locking. Angle fracture fixation showed maximum deflection of 5.93459mm with Orthomax conventional and least of about 3.00287mm with Synthes locking plates. Maximum stress over bone was more of about 379.81 Mpa for Orthomax conventional and least of about 309.63 MPa for Synthes locking plates. Stress over plate was more of about 2114.62 MPa in Orthomax conventional and least of about 833.457 MPa in Synthes locking. Fracture gap was more of about 2.2708mm in Orthomax conventional and least of about 1.86241mm with Leforte locking.

CONCLUSION: Consecutive rapid failure of the miniplates could not be prevented when the angle and symphysis region are loaded with vertical bite forces. The more stable plate is Synthes locking plate followed by Leforte locking plate for the symphysis region and angle region. The static yield limit of titanium exceeds, when geometry and dimension of the miniplates get altered, while adapted to angle and symphysis region. Hence, the dimensional changes in the holes of miniplates occurring during adaptation of the plate to the fracture site are also a factor to be considered for stability of the plate.

KEY WORDS: Biomechanics, mandibular symphysis and angle fractures, fracture fixation, conventional miniplates, locking miniplates, Three Dimensional Finite Element Analysis, bite force.

LIST OF ABBREVIATIONS

CAD	Computer Aided Design
CT	Computed Tomography
2D	Two Dimensional
3D	Three Dimensional
FEA	Finite Element Analysis
FEM	Finite Element Model
viz	namely
TMJ	Temporomandibular joint
IMF	Inter Maxillary Fixation
N	Newton
N/mm^2	Newton per millimeter squared
MPa	Mega Pascals
GPa	Giga Pascals
mm	millimetre
cm	centimetre
Kg	Kilogram
DICOM	Digital Imaging and Communications in Medicine
IGES	Initial Graphics Exchange Specification
Ti	Titanium
SRIF	Semi Rigid Internal Fixation
VMS	Vision Measuring System
avg	Average
RTA	Road Traffic Accident

CONTENTS

INTRODUCTION

The continued interest on biomaterials, motivated Physicians and Maxillofacial surgeons pay a lot of attention towards investigation of mechanical properties of bone tissue and other biocompatible materials. The first investigation on bone mechanical properties commenced in the nineteenth century.[1] The Rational structure of bone and the effective resistance of bone tissue to mechanical loads can be used as a prototype for development of new composite materials and designs that becomes compatible and economical to be used in medical practice.[2,3,4,5,6,7,8,9]

Mandibular fractures are the highest of all the facial bone fractures in any maxillofacial trauma situation.[5,6] Mandibular fracture affects healthy breathing, food consumption and clear conversation. The above, when not handled with care, causes serious deformities affecting the normal life of the patient and leading to secondary correction.

Biomechanical study revealed that masticatory muscles produce tension at the upper border and compression at lower border of mandible.[10,11,12,13,14,15] Torsional forces are produced from incisors to canines i.e., in the chin area.[12,16,17] Miniplate osteosynthesis, introduced by Michelet et al in 1973[18] and further developed by Champy et al in 1978[1], has become the standard treatment of mandibular fractures. Hence the era of miniplates for osteosynthesis by means of load sharing developed.[1,16,20,21,22] They described monocortical tension banding osteosynthesis, neutralizes distraction and torsion during physiologic stress.[12] Despite the fact that mechanical properties of bone tissue have been investigated for long time, behaviour of bone tissue material in fracture osteosynthesis with Semi Rigid Internal Fixation

(SRIF) method is not sufficiently explored. One of the types of SRIF is Miniplates osteosynthesis. This helps early mobilization of the mandible for normal functioning and also avoids wires /splints/long term arch bars (Nonrigid fixation technique) for stability of the fractured site for osteosynthesis. Moreover Nonrigid fixation technique is old and has enough number of complications to be specified which could be overcome by SRIF.[20]

Advantages of SRIF:

- Increased fractured segment stability (comparatively)
- Prevents long-term intermaxillary fixation
- Enhances fast bony osteosynthesis.

MOTIVATION OF THE RESEARCH

During normal activities, fractured mandibular bone is exposed to cyclic loading.[21] Highest stress arise from the cyclic loading action when applied vertically on to the occlusal and incisal region of the mandibular teeth. This may compromise the initial fixation of the miniplates which leads to bone tissue resorption and loss of contact between miniplates and bone tissue. Adequate treatment choices prevent the consequences of the given stress. The design and dimension of the miniplates plays a major role in the stress distribution over the parabolic (horse shoe shape) mandibular bone.[22]

The adaptability of miniplates depends on factors such as the material properties, design, and biocompatibility. Biocompatibility includes mechanical properties such as strength, stiffness and viscoelasticity.[16,23,24] Incorrect choice of miniplate design, may induce additional stress concentration over material and bone tissue structure. The assessment of stress and deformation, that induce cyclic loading, is essential to predict the risk of failure.[19,25,26]

Manual experimentation needs physical models, which are inconvenient, costly and difficult to simulate normality in certain situations. This paved the way for development of virtual analysis using three dimensional (3D) Finite Element Modelling (FEM) which is more convenient and solves all complex situations in reproduction.[10,27] The Finite Element Analysis (FEA) method is widely accepted replacement for photo elasticity tests, as it is one of the most practical and reliable methods. FEA numerically analyses mechanical behaviour of biological objects.

Within this thesis, the focus will be on simulating the fixation of angle and symphysis fractures of dentulous mandible using 2.0mm titanium miniplates both conventional and locking plates along the Champy's line of osteosynthesis. Loading cases can range from fast movement and small forces when talking, to slow or no movement and large forces during mastication. The aim of the study is to evaluate the stability, flexural and compressive strength and deflection of miniplate system, on occlusal forces in symphysis and angle fractures of mandible elicited on human cadaveric mandible model with Finite Element Analysis (FEA) by mechanical application of vertical load.

AIMS & OBJECTIVES

The purpose of the Finite Element study is

1. To evaluate dimensional changes, stability and deflection of miniplates after adapting to the symphysis and angle region.
2. To evaluate mechanical stress over bone.
3. To evaluate maximum stress bearing area over miniplates.
4. To evaluate the distance between the reduced and fixed fractured segments (fracture gap) after vertical load application.
5. To evaluate direction of displacement of the fractured segments.

REVIEW OF LITERATURE

Michelet F.X. etal (1973)[18] did an analysis on 300 cases involving fractures of the facial bones that were treated using malleable non compression miniplates made of Vitallium. Monocortical screws were used for fixation and a 3 year follow up was done. They concluded that the miniplates were fixed through adequate intraoral access and there was perfect adaptation of the fractured segments. The complications were also reduced to 5 % and also the requirement for intermaxillary fixation was reduced.

James P. Ralph and Angelo A. Caputo (1975)[5] prepared replicas of dentate human mandible in photoelastic liquid epoxy material to study the stresses that develop within the structure of mandible in response to the application of various loads. Occlusal loading was simulated and the stresses generated within the models were examined by 3D photo elastic stress analysis in three different loading conditions. It provided a means for demonstrating and analyzing the stresses likely to be imposed on the structure of mandible by various clinical procedures.

Maxime Champy etal (1978)[1] studied a series of 183 cases with fracture of the mandible treated with modified Michelet's Technique and analysed in a 5 year follow up study. The osteosynthesis was accomplished by monocortical, juxta alveolar and subapical osteosynthesis done by intraoral approach without intermaxillary fixation. The parameters studied were stress distribution, photo elasticity and movements at the fracture site. The use of a too rigid lower border plate was inadvisable because it resulted in the shield effect. The authors concluded that the modified Michelet technique was the most successful in the management of mandibular fractures except in cases of pre-existing infection as a natural strain of compression was present in the lower border, compression plating was no longer advisable.

Franco Mongini etal (1979)[3] studied relationship between the shape and structure of the mandible & their stress pattern distribution when occlusal loads were simulated. 10 human skulls 4 males and 6 females aged 22 to 53 were evaluated. Right & left lateral cephalographs were taken to know the internal density of the mandibles and 500 points were graphically plotted using software. Occlusal loads were simulated & isoclinics were in plane polarized light & relation between the occlusal load & stress pattern recorded on the brass replica of the model. Results showed stress pattern prolongation over the angle, ramus & condyle predominantly. No correlation of the orientation pattern with sex & age were observed.

Proffit WR, Fields HW and Nixon WL (1983)[59] studied accurate measurement of occlusal forces in humans and provided data for normal and long face adults. The study used both quartz and foil based piezoelectric force transducers to evaluate occlusal forces during swallow, simulated chewing and maximum effort in 14 long face and 21 normal individuals at 2.5mm and 6.0 mm molar separation. The results of the study showed that the occlusal forces in normal and long-face adults at 2.5mm and 6.0mm molar separation during swallowing was 2.9kg, 4.8kg, 1.1kg, 1.8kg respectively, chewing was 13.5 kg, 16.2kg, 4.2kg, 4.8kg respectively and the mean bite force was 31.0 kg, 35.6kg, 11.2kg and 15.5kg respectively. The study concluded that the long-face individuals had significantly less Occlusal force than do individuals with normal vertical facial dimensions.

Cawood J. L. (1985)[15] evaluated the effectiveness of mini plate osteosynthesis by comparing mini plate osteosynthesis with direct fixation and intermaxillary fixation for six weeks for the management of mandibular fractures. The sample size in each group were 50 cases. The mini plate osteosynthesis resulted in sulcus loss in 16% of the patients, wound dehiscence in 12%of the patients and malocclusion in 8 % of the patients, infection in 4 % of the patients and sensory disturbance in 4% of the patients. Though complication occurred, plating resulted in earlier healing and jaw opening. Intermaxillary fixation could be used as a supplement in cases of comminuted mandibular fractures.

Charles H. Gibbs etal (1986)[58] designed & developed Gnathodynamometer for analysing bite forces of human. It consisted of two stainless steel plates separated by a steel sphere that balanced biting force between right and left sides. Gnathodynamometer was placed bilaterally between premolars, first molars, and second molars. The interocclusal separation between the posterior teeth was approximately 12 mm. The greatest bite strength, 975 lbs (443 kg), was recorded from a 37-year-old man. Bite strength of 975 lbs (443 kg) was found for one subject in this study which was more than twice any previously reported value and 2.8 times more than the value that was reported for Eskimo. Bite strength in some bruxer-clenchers was six times more than that of a nonbruxer.

Imre Karasz etal (1986)[4] Did Photoelastic stress analysis to analyse 3 different methods of osteosynthesis a) Figure-of-8 wiring (b) Eccentric Dynamic Compression Plates (EDCP) and (c) Champy's plate. An 8mm thick Araldite B photoelastic material was used to fabricate an average mandible model fixed with 3 different osteosynthesis and the mechanical response was analyzed under two loading forces representing the normal masticatory forces. The results of the study indicated that with Figure-of-8 wiring only a minimum compression effect was seen and in the alveolar area with full load, the gap was opened and the pressure increased. The usage of EDCP plate resulted in pressure in the alveolar region as a result of full loading.

The Champy plate served in the juxta-alveolar region created clearly visible compression stresses in the mandibular base on application of loads. The study concluded that the Champy method was the most resistant to vertically bending forces and satisfied the final requirements for a functionally stable osteosynthesis.

Ashman R.B. & Bushirk W.C.V. (1987)[6] conducted the study to understand how the internal forces were distributed through the mandible and how the mandible deformed as a consequence of these internal forces. Using a continuous wave ultrasound technique the elastic components of ten small specimens of about 3+5+5mm were determined. The technique involved passing a continuous ultrasonic wave through a specimen and the time delay corresponding to the propagation of the wave through the thickness of the specimen by making a phase comparison of the signal before and after its transmission was made. Linear elastic wave theory showed relationship between various velocities through the specimen and its elastic properties. The study indicated that human mandibular bone was elastically homogenous but anisotropic. Elastically it behaved like a long bone bent into the shape of a horseshoe (parabola).

Rozema F.R. etal (1992)[40] used several types of fracture fixation devices to analyse the strength of the osteosynthesis plates and screws for undisturbed fracture healing. The aim of the study was to analyse the possibility for application of resorbable plate osteosynthesis to determine the optimal site and direction of osteosynthesis in individual cases. A biomechanical integrated model of the mandibular system was used in the study. The 3D location of the fracture and the anatomical location of the osteosynthesis plates and screws were indicated on the biomechanical model. The model was subjected to occlusal and muscular forces. The displacements and the rotations across the fractures at a given position of one or more bone plates under the occlusal load were calculated. The study evaluated the point at which the lowest maximal bone plate strain occurred and the site at which all forces and torques cancel each other. The study concluded that the type of fracture (serrated or smooth), the strength and shape of the osteosynthesis plates influenced the choice and positioning of bone plates and screws used for fixation of mandibular fractures.

Luis A. Passeri etal (1993)[20] analysed the complication rates associated with closed reduction and open reduction using non-rigid means of fixation in 96 patients with mandibular angle fractures. The results of the study indicated rate of infection was 13%. Four cases had combined infection with malunion and malocclusion. The study concluded that the mandibular angle fractures were associated with a significant number of complications, regardless of the method of treatment.

Gregory S.Tate etal (1994)[30] evaluated the ability of patients with fractures of the mandibular angle to generate bite forces after surgical treatment. 35 male patients were treated for mandibular angle fractures with rigid internal fixation using a custom made bite force gauge. The results indicated that in week 1 through 6, the average incisor bite force was only 6.4 kPa compared with 12.3 kPa after 6 weeks, and 15.4 kPa in control group. There was no significant difference after the 6th post- op week. Like the incisior bite force, in weeks 1 through 6, the average right and left molar bite forces were 13.8 and 12.8 kPa respectively, compared with 25.3 and 26.0 kPa after 6th post-op week, and 48.2 and 49.3 kPa in control group. The average bite force on the fractured and nonfracture sides were 13 kPa and 18 kPa, before 6 weeks and 27kg after 6 weeks respectively. Thus the duration for regaining normal bite forces after fracture fixation was found to be reduced.

Byung Ho Choi etal (1995)[60] did clinical and in vitro evaluation of mandibular angle fracture fixation with the two-miniplate system by placing one plate at the superior border and a second plate is applied at the inferior border of the buccal cortex compared with single miniplate placed according to Champy's method. Opening of fracture gap in angle fracture was present in Champy's method and no visible opening of gap in angle fracture with two-miniplate-fixation technique. This was confirmed when 40 mandibular angle fractures were treated with the two-miniplate fixation technique. Bone healing took place in all cases without evidence of osteomyelitis. Hence two miniplate fixation was superior to conventional technique in stability and resistance in angle fracture.

Osborn J. W. & Baragar F. A. (1995)[54] predicted the pattern of human masticatory muscle activity during clenching where symmetrical vertical bite forces were simulated in the 3D model of jaw. A total of 26 muscle elements like the upper and lower attachments of the muscles of mastication had been measured on a single human skull and divided into thirteen independent units on each side. For symmetrical biting, the model minimized the sum of the muscle forces used to produce a given bite force and activated muscles in such a way which corresponded well with previous observations on human subjects. The muscles were divided into two viz power muscles and control muscles. Power muscles were masseter, medial pterygoid and temporalis which produced the bite forces but tends to displace the condyle up or down the articular eminence. This displacement was prevented by control muscles (oblique temporalis and lateral pterygoid) which had very poor moment arms for generating usual bite forces but were efficient for preventing condylar slide. Hence during vertical occlusal load / clenching the condyle remains static.

James W. DeVocht etal (1996)[55] developed a two dimensional finite element model to simulate in-vivo biomechanics of TMJ over the range of normal motion. Model was developed using ABAQUS software with slide line elements that allowed large displacements and arbitrary contact of surfaces. The three main components of the model were the mandibular condyle, articular disc, and glenoid fossa region of the temporal bone, which were all modeled as deformable bodies. Continuous motion was simulated by doing a static analysis for each of many small steps. A parametric study was performed by determining the maximum stress in each of the three main components as a function of the elasticity of the articular disc. This model suggested that muscle contraction was not required to maintain proper disc position. Normal motion resulted in relatively high stresses deep in the glenoid fossa.

Richard H. Haug, J. Edward Barber And Robert Reifeis (1996)[51] compared the conventional plating technique for mandibular angle fracture with two biomechanically dissimilar techniques in their abilities to resist vertical loads similar to masticatory forces in vitro. Conventional treatment of mandibular angle fractures used thinner and smaller tension band at superior border and thicker and larger stabilization plate at inferior border. Non-traditional mandibular angle fracture plating group used thicker tension band at thicker superior border and thinner stabilization plate at thinner inferior border. Two miniplate group used two thin plates: one as stabilization plate and other as tension band. 100% of the failures occurred with monocortical screws at the tension band/synthetic bone interface of the superior border. In addition, 80% of the failures occurred at the two most anteriorly positioned screws.

Righi E. etal (1996)[22] experimented ten segments of bovine scapula bone. Five of which fixed with 4-hole titanium miniplate and other five fixed with 6-hole Double Y plate and adapted to a tension test machine. Shear tests were recorded. On the basis of the test results, two simple computer models were developed. No significant difference in stress pattern was evident between the mechanical and computed tests. The most critical sections were located near the hole proximal to the osteotomy and the microscopic findings confirmed this. This suggested that straight miniplate design would provide sufficient stability and a high degree of anatomical adjustment of the system.

Tams J. etal (1996)[11] formulated criteria for the mechanical properties of the osteosynthesis systems. The loads across the fracture and their influence over the fracture patterns were analysed. In a 3-D mandibular model, bending, torsion and shear forces for fractures located in the angle, anterior body, posterior body, canine and symphysis region were calculated with simulated muscle and bite forces. The results of the study indicated that the angle and posterior body fractures had high positive bending movements, small torsion movements and high shear forces. The anterior body, canine and symphysis fractures had high negative bending moments and higher than the maximum torsion movements. The study concluded that mandibular fractures can be roughly divided into two groups which had similar patterns of loads across the fracture. Angle and posterior body form a group and the anterior body, canine and symphysis fracture form another group. Neutralization of these forces was influenced by the mechanics at the fracture site and the mechanical properties of the implants. Hence the influence of the fracture characteristics and the mechanical property of the implant should also be considered before formulating a criteria for the number of plates and position on the bone.

Joerg M. Wittenberg etal (1997)[2] evaluated the effectiveness of fixation devices of simulated angle fractures in sheep mandible. The angle fractures were stabilized by Leibinger 3D plates, Synthes 8 hole mesh plate and Synthes reconstruction plate with 2.0mm & 2.4mm mono & bicortical screws. Each mandible was tested for bending and Leibinger 3D plates showed deformation in bending at force greater than 230 N. None of the plates showed failure in bone screw interface.

Tams J. etal (1997)[12] compared bending and torsion movements across mandibular fractures for different positions of bite points and different sites of fracture. The study was conducted on three identical resin duplicated mandible models on which angle, body and symphyseal fractures were simulated and fixed using a polyurethane plate with 2.0mm metallic self-tapping screws with an interfragmentary gap of 3mm. The muscle and bite forces were simulated. The results of the study indicated that the Body and symphyseal fractures had smaller positive bending moments and higher negative bending moments with higher torsion moments. Hence this required fixation with either a single plate or two plates which equally gave good clinical results. The study also evaluated the transmission of load patterns across the smooth and serrated fracture. For a smooth fracture, interfragmentary stability was less and hence bone plates had to carry a larger part of the loads, demanding more of the fixation device.

James W. Sikes etal (1998)[53] compared the clinical advantages and the ability to achieve stability with fewer number of screws per bony segment between locking head screws and conventional screws. Two-dimensional beam mechanics technique was used. Adult bovine ribs and the Instron machine were used to develop a load-displacement curve up to 150 N for each specimen. An osteotomy was then created and the segments were reduced with preload (fracture model) or with a 1-cm defect (reconstruction model), and plated using the Synthes locking-head plate with either two or four bicortical locking-head (4.0mm) or conventional (2.7mm) screws per segment. The fixed ribs were loaded to 150 N and the displacement was recorded. When two screws were used locking screws showed increased resistance to displacement and when four screws were used there was no significant difference between locking-head and conventional screw types in either model.

Carl E. Misch, Zhimin Qu & Martha W. Bidez (1999)[8] studied nine fresh-frozen human mandibles between the ages of 56-90 years. The anterior (incisors and canine), middle (Premolars), and distal (molars) sections were cut. Each region was then further sectioned horizontally. Cortical bone superior and inferior to the trabecular bone were removed leaving two plates with a thickness of 5 mm. The trabecular bone specimen (side-constrained) was subjected to a non-destructive compressions load (rate of 0.015) after which cylindrical trabecular specimens (Unconstrained) were sectioned in the same site with an inner diameter of 5.0 mm (5mm diameter 5mm height) and subjected to similar compressive loading as the side constrained specimens. This was done to determine any differences in elastic modulus based on constraint condition imposed by the buccal and lingual cortical plates. The study concluded that the anterior mandible had a greater trabecular bone density which also correlated to a greater compressive strength and elastic modulus than in other regions.

Edward Ellis III (1999)[14] evaluated the efficacy, least complication that could occur and simplicity of the plating techniques, types of plates, number of them used and their approach in mandibular angle fracture. The techniques included: 1) closed reduction or intraoral open reduction and non-rigid fixation; 2) extraoral open reduction and internal fixation with an AO/ASIF reconstruction bone plate; 3) intraoral open reduction and internal fixation using a solitary lag screw; 4) intraoral open reduction and internal fixation using two 2.0 mm mini-dynamic compression plates; 5) intraoral open reduction and internal fixation using two 2.4 mm mandibular dynamic compression plates; 6) intraoral open reduction and internal fixation using two non-compression miniplates; 7) intraoral open reduction and internal fixation using a single non-compression miniplate; and 8) intraoral open reduction and internal fixation using a single malleable non-compression miniplate. Intraoral open reduction and internal fixation using a single miniplates were associated with the fewest complications and the easiest with reliable stability.

Peter Maurer, Siegfried Holweg & Johannes Schubert (1999)[38] developed a three dimensional Finite element model of the mandible and studied the biomechanical loads of osteosynthesis screws in bilateral sagittal osteotomy. When bite forces were applied, the most stable configuration was found to be a triangular one. A mini screw of 2.0 mm diameter can provide sufficient stability at the osteotomy site after ramus split osteotomy. Even screws with a diameter of 1.5 mm would withstand forces up to 89.5 N, which would not normally be reached by patients after ramus split osteotomy in the early period of healing. The tensile stress distribution in the lowest screw subjected to the highest loads. The analysis of the tensile stress in the 1.5 mm screw revealed that the linear configuration only tolerates lower chewing force. Forces exerted by patients after bilateral ramus split osteotomy did not exceed these values.

Dirk Vollmer etal (2000)[7] applied well defined forces to human mandible in an experimental set-up. The resulting strains were measured with strain gauges adhered to different anatomic positions. A 3-D CT of this mandible was performed before these measurements. Based on the CT data, a voxel orientated FE mesh was generated. Any changes in the mechanical behavior and measurements were measured on the same specimen three times within 4 weeks by applying 150 N. All the strains at all measured points were still within the linear elastic region. It was concluded that FE model was a valid, accurate, non-invasive method to predict different parameters of the complex biomechanical behaviour of human mandibles with respect to load transfer, stress distribution and displacements.

Wieslaw W. Chladek etal (2000)[57] did a study using FEA to evaluate the force share exerted by each individual muscle and amount of forces necessary to ensure the stability of the mandible. Three dimensional models were exerted with a force of assumed constant value & qualitative & quantitative evaluation of phenomenon done in right & left sided canine, second molar and anteriors. Without the support of one joint the mandible is maintained in equilibrium with the change of forces from lateral pterygoid muscle and temporal muscle to a lesser degree without the influence of masseter & medial pterygoid muscles on stabilization.

James W DeVocht etal (2001)[56] performed a study to show that the previously reported FEM of TMJ provided results that were reasonable for approximation of the actual physical situation. The study was done on a 2-D FE model of the TMJ in which the mandibular condyle was depicted to rest in the glenoid fosse initially with articular disc in-between. Normal mouth opening was simulated with mandibular condyle being rotated and translated down while applying a vertical force to the mandible. The stresses calculated were compared with an experimental model of a human cadaver mandible. The results of the study indicated that the FE method provided a reasonable representation of the biomechanics of the TMJ.

Gerlach K. L. & Schwarz A. (2002)[13] evaluated the bite forces for 22 patients between 24 to 38 years of age who underwent miniplate osteosynthesis according to Champy's principle for mandibular angle fracture. Bite forces at incisors, canines and molars was carried out 1 to 6 weeks following the treatment and also in 15 control group. The members of both groups had complete dentition with or without 3rd molars, only minimal dental restorations, no sensitivity to percussion on teeth to be tested and had agreed to participate in this study. Bite forces between the molars in the study amounted to 90N in 1 week and 148 N at 6 weeks postoperatively.

Cox T., Kohn M.W. And Impelluso T. (2003)[10] conducted FE study to assess whether rigid fixation by resorbable polymer plates and screws could provide the required stiffness and strength for a typical mandibular angle fracture. For the study, an existing computer model of the human mandible was procured from viewpoint software in which the interface between cortical and trabecular bone was determined from CT scans using SIMPLANT software. The trabecular bone was assumed to be isotropic with an average tensile modulus of 56 MPa and a Poisson ratio of 0.3. An angle fracture with an interfragmentary gap of 2mm was simulated and a standard, double plating technique with 2.4mm diameter monocortical screws on the superior plate and 2.4mm diameter biotical screws on the inferior plate was selected, simulated on two different FEA meshes, one set for resorbable Osteosynthesis and other for titanium plates and screws. After assembly of all the components, the model was subjected to bite and muscle forces in which plates were assumed not to receive or transmit any force directly from the bone segments rather the chain of force transfer was defined as progressing from bone to screw, screw to plate and finally returning via the screws back to the bone. Finally stresses and bone interfragmentary displacements in each mandible model and in the fixation devices were assessed for the appropriate bite and muscle forces. The calculated stresses and displacements were then compared with yield strength of each material and with each other. Finally overall stress patterns in the fractured mandibles were compared with each other. The results indicated that mandibles, fixed with either titanium or resorbable materials showed nearly identical stress patterns with maximum displacement less than 150 um to meet the currently established norms for fracture immobility. The study concluded that resorbable polymer plates and screws were of adequate rigidity for fixation of mandibular angle fractures.

Gabrielle M.A.C. et al (2003)[47] conducted a study on 191 cases. The samples were mostly males a common site of fracture was angle of mandible. The cases were inspected for infection, malunion, and fibrous reunion and missing of follow ups. The fixation of mandibular fracture with 2.0mm miniplates has similar incidence of complications such as inferior alveolar nerve paraesthesia, temporary mild deficit of the marginal mandibular branch, hypertrophic scars in extra oral approach, occlusal alterations, facial asymmetry, malunion, fibrous union, and condylar resorption as that of rigid methods of fixation.

Jose R. Fernandez etal (2003)[39] created a Finite element model of the human cadaveric mandible and 3 situations were simulated and tested for maximum Von Mises stress. Two compressive lateral forces were applied first to the healthy mandible, second to the simulated symphysis fractured mandible fixed with two miniplates one at the tension border and other at the compression border and finally to an osseo-integrated mandibular model. Von Mises stress was high in the symphysis region in all the 3 models but fracture gap was within limits in fractured model and no gap was found in the osseo-integrated model. Thus miniplates were a better option in case of stability.

Kay-Uwe Feller etal (2003)[26] - computed masticatory load on different osteosynthesis plates in a FEA model and determined whether miniplates were sufficiently stable for application at the mandibular angle. A FEM of human mandible was developed using 20 mandibles of human cadavers and the directions and volumes of muscle vectors were simulated on the basis of studies by Schumacher (1961). An angle fracture was simulated and fixed with miniplate (1.0mm thickness) or a module plate (1.5mm thickness). A 30% reduction in masticatory load (Studies by Tate et al) was simulated on the osteosynthetic devices approx. 2cm away from the fracture gap and their stability was computed. In the second part of the study data from 277 patients with 293 fractures of the mandibular angle were evaluated retrospectively. The results of the study indicated that both the 1.0mm miniplate and the 2.3mm module plate were sufficiently stable. The study concluded that a single miniplate was sufficiently stable to withstand the reduced masticatory forces in angle fracture cases.

Lettry S. etal (2003)[9] studied regional distribution of the mechanical properties in the cortical bone of the mandible. Investigation was done simultaneously in both regional and directional variations for mechanical properties. Computed Tomography (CT) numbers of the fresh mandible bone provided a noninvasive method for determining the bone quality for designers of dental implants. Using a three-point bending test the regional variation of Young's modulus of bone in the human mandible was determined from five fresh specimens fully dentulous, some partly edentulous and some completely edentulous. A pattern of modulus distribution was evident in the mandibles. Directional variations of the modulus for the mandible were due to the anisotropic nature of bone.

Torreira G.M. & Fernanedz R.Z. (2004)[28] studied a 3-D FE mandible model to determine the patterns of biomechanical responses in order to identify the highest stress zones where fractures might occur under standardized trauma conditions. The FE mesh of human mandible was achieved by means of a scanner with laser technology digibiotics II 3D and the software AutoDesk 3D studios. The model was assumed to be isotropic with homogenous elastic properties. This was then subjected to a load of 107N/m2 in the symphysis region and in the body of the mandible. The resulting deformations of the mandible were studied 1 sec after the blow was produced. Stresses and displacements in the mandible were calculated by Fem analysis using 3D Von Mises stress norm. Following a blow to the symphysis region, maximum stress areas were located at the symphysis, retromolar and condylar re gions. On a blow to the mandibular body, the maximum stress were located at the contralateral angle, ipsilateral body and the ipsilateral condylar neck regions.

Cheng-Jen Chuong etal (2005)[49] did a study using FEA to compare mechanical stress after stabilizing the BSSRO by using 2 different techniques – 3 bicortical titanium screws and curved titanium plate with 4 monocortical screws. Deflection and stability of the bicortical titanium screws were much better when compared to miniplate system and the deflection was more at the central incisor region in both the techniques during the bilateral load application. Von Mises stress on the bone was high in the vicinity of screws in the unilateral 1kg compressive load in both the techniques. Miniplate system had high mechanical stresses when compared to the inverted L configuration of the bicortical screws both in torsion and bending.

Erkmen E. etal (2005)[35] evaluated the mechanical behaviour of different fixation methods used in 5mm advancement BSSO using finite Element Analysis (FEA) with occlusal load over molars and premolars. Four techniques, where 3 lag screws in triangular configuration, 2 lag screws parallel to each other, 2 parallely placed six hole with gap miniplates with monocortical screws and single oblique configuration miniplate were fixed. Complex mechanical stress was simulated with 500 N posterior occlusal load on distal segments. The use of 2.0 mm lag screws placed in a triangular configuration surgery provided sufficient stability with any rotational movement and less stress fields at the osteotomy site, when compared with the conventional and locking miniplates fixation methods.

Hideki Takada etal (2006)[23] researched that the factors related to bone fractures include the amount and direction of load, biomechanical properties of bone and anatomical properties. The presence of increased risk for an angle fracture was associated with presence of an impacted third molar. A finite element model from distal to 2^{nd} molar to ramus region was created one with impacted third molar & other without third molar. When load was applied to the angle, the major area of destruction was in the angle region. A load of 2000N was perpendicularly applied to the buccal surface of the body of mandible. Histogram was prepared for each element to identify the stress distribution. The Volume of Interest (VOI) analysed in the Von Mises equivalent Stress distribution. Thus, bone density determined the mechanical strength of the bone. In the mandible with third molar, the stress was concentrated more in the root apex. The presence of a third molar thus changed the concentration and transmission of stress in the mandible. This increased the risk of an angle fracture.

Ziebowicz A. & Marciniak J. (2006)[32] analysed a three-dimensional finite element model to simulate the evolution of the displacement & stress fields of fractured mandibular angle after reduction with miniplates osteosynthesis. The system was designed permitting the muscle forces exerted on the model numerically. Young modulus and Poisson's ratio were given as material properties. The gap of the fracture under the applied forces did not exceed 1mm. Maximum dislocations in two miniplate osteosynthesis system did not exceed 0.051mm. The investigations showed that two-miniplate stabilizing system ensured appropriate stabilization.

Hasan Husnu Korkmaz (2007)[21] formulated a 3D Finite Element Model of fractured mandible fixed with miniplates system which was tested for biomechanical stabilization after application of occlusal loads. 19 different miniplate configuration & techniques included X plates, 90 degree L-plates, T plates, 2-hole with gap in combination with 4 hole without gap, two 4-hole with gap and single 4 hole with gap titanium plates were evaluated for torsion, bending, stability and deflection. The most appropriate arrangement was 4-hole miniplate on the superior border and a 2-hole miniplate in the inferior position. Use of an X type miniplate was considered as an alternative to two 4 hole with gap type miniplates which provided sufficient stability.

Hamdi Arbag etal (2008)[27] performed FEA to identify the suitable shape and fixation technique for mandibular corpus fractures. 14 different techniques of fixation of 4 hole with gap miniplates either single or double, 6 hole without gap miniplates either single or double, 90degree L plate and T plates were fixed at different levels and evaluated for displacement and stresses in the titanium miniplates for these configurations. Results indicated that the use of 2 straight miniplates was more rigid than other fixation types and fracture mobility was approximately equal to or less than the limit. This study concluded if any displacement more than 150 micrometer was present then that technique was less stable.

Scott T. Lovald, Jon D. Wagner & Bret Baack (2009)[19] created a finite element model of a fractured body of human mandible using tomography scans. Material properties were assigned to the cortical bone, cancellous bone, and dental region. The authors compared the efficacy of new internal mini locking plates and conventional miniplates for stability. Osteosynthesis and stability of fixation was preferred in a 3D in vitro model in which functional load was stimulated. On comparing different osteosynthesis technique, locking miniplates system showed less torsion and gapping of the bone to conventional miniplate system. Stability was more in interflex II miniplates. Locking miniplates with screws and cortical bone formed a frame work which increased the stability. Hence it was concluded that mini locking system were proved efficient in osteosynthesis.

Tsutomu Sugiura etal (2009)[61] investigated the biomechanical behaviour of the miniplate osteosynthesis using FEA. 3 different atrophied and non-atrophied mandibular models were prepared each with 30mm, 20mm, 15mm and 10mm height and fixed with single and double 4 hole with gap 2.0mm titanium miniplates & 7mm long screws. Single miniplate area of fixation was at the superior border of the mandible. No noticeable difference was identified in the compressive stress level in all the models. The use of double miniplate showed a great influence on Von Mises stress reduction in the miniplates, minimal displacement between the interfragments and minimal gap due to torsion. Thus double miniplate was more reliable method in treating anterior fractures.

Baohui Ji etal (2010)[34] constructed two 3D FE models which simulated symphysis fracture of mandible fixed with single 2.0mm miniplate in the centre and double miniplate in parallel fashion according to Champy. The study evaluated the stress distribution and stress shielding effect of titanium miniplates used for the treatment of symphyseal fractures. Ratios of stress shielding effect with the lower miniplates in technique 2 were much higher than the upper miniplates and the miniplates in technique 1 during all conditions, and that value of the lower miniplate gained a maximum value of 83.34% during left unilateral molar clenching. The stress areas were concentrated on the central section of the miniplates. However, the stress distribution varied with masticatory conditions. The study demonstrated that miniplate stress distribution and stress shielding effect ratio were affected not only by the way in which the mandible was loaded but also by the number of the miniplates fixing the fracture.

Hang Wang etal (2010)[17] studied Von Mises stress distribution in the symphysis fracture of human mandible fixed with single 4 hole miniplate in the middle and two 4 hole miniplate in the tension and compression border using Finite Element Analysis. Two basic loading condition like left unilateral molar clenching and bilateral intercuspal clenching were simulated. The maximum stress occurred at the biting point. Two miniplates have a biomechanical advantage over 1 miniplate in stabilization.

Ribeiro-Junior P. D. etal (2010)[44] analysed the influence of the type of miniplate and number of screws on the stability and resistance used for mandibular angle fracture fixation. Sixty polyurethane hemimandibles assigned in four groups were fixed using 2.0 mm 4 hole or 7 hole miniplates and 2.0 mm x 6 mm monocortical screws either locking or conventional. The hemimandibles were loaded in compressive strength until a 4 mm displacement occurred between the segments, vertically or horizontally. Locking plate and screw system provided significantly greater resistance to displacement and stability than conventional ones.

Atson Carlos de Souza Fernandes etal (2011)[24] examined 100 adult human cadaveric dentate hemimandibles for thickness of the cortical bone between canine, first and second premolars for insertion of monocortical screws to place miniplates during mandibular fractures. This was using CT imaging and assessed the precision of the dimension. Assessment made by CT was compared with the same measurement measured directly with calipers. There was no significant difference between the two methods. The vestibular cortical bone was less than 3.0mm thick and inter-radicular distance between canine and premolars was more than 2mm. In the mental foramen region cortical bone thickness was less than 3mm. Thus the inter-radicular distance suggested minimal risk of radicular injury on miniscrew insertion between alveolar structures.

Julie Kimsal etal (2011)[42] analysed 3 separate scenarios using Finite Element Analysis. Angle fracture was simulated and fixed with 1) a 6-hole bicortical angle compression plate at the inferior border of the fracture; 2) a tension band alone at the superior border of the fracture; and 3) the combination of the plates in the previous 2 scenarios. The dual plate model incurred the lowest Von Mises stresses in the plates and the lowest principal strain in the callus. A single tension band on the superior border provided more angle fracture stability than a single bicortical plate placed inferiorly and provided sufficient stability compared to combination plate fixation scheme.

Prabakhar C. etal (2011)[45] conducted a study to prove the efficacy of 2mm locking in stabilizing fractured mandible without maxillomandibular fixation. It was prospective study involving 20 patients who had various degrees of mandibular fractures. The selected cases were fixed with open reduction and internal fixation using 2mm locking plates in different regions of fractured mandible. The cases with pre-operative infection were excluded and those with systemic complications were also excluded. In this study conducted in a randomized walk in patients concluded that fixation of mandibular fractures using 2mm locking miniplates proved efficient enough than maxillomandibular fixation.

Goyal M. etal (2012)[50] conducted a study on 30 patients with mandibular fractures in which two groups were divided each containing 15 members. Group A and Group B were treated with monocortical and lag screws respectively and follow up were made at 3, 6, 12, 24 weeks. Rigid internal fixation provided by lag screw technique for anterior mandibular fracture was advantageous over conventional bone plating in stability.

Gupta A etal (2012)[31] conducted a study to compare 1 micro plate combined with 1 miniplate with standard 2 miniplate in their efficacy and clinical stabilization. Bite force measurement was done in a sample of 10 in group A treated by 1microplate and 1miniplate and 10 in group B treated by standard 2 plates. The bite force measurement pre-operatively and post-operatively revealed that using a 1miniplate and 1 micro plate in the management of mandibular fractures was stable and adequately efficient to withstand masticatory loads and torsional forces acting in the anterior region of the mandible.

Andre Vajgel etal (2013)[25] analysed three dimensional finite element models for biomechanical stability of 2.0mm locking miniplate fixation in mandibular fractures. The models were divided into 4 groups according to plate thickness (1.0, 1.5, 2.0, and 2.5 mm). Fractures were simulated in left mandibular bodies, and 3 locking screws were used on each side of each fracture for fixation. Bite forces were simulated in the incisor and molar regions of the mandibles in finite element models. Von Mises stress was high during simulated bites in the molar region for plates with thicknesses of 1.0 mm. Plate tension values were below the level required for permanent deformation or fracture in all models. The 2.5-mm-thick plate presented better biomechanical performance than all other plates. The 2.0-mm-thick plate also showed adequate safety limits.

Bipin S. Sadhwani etal (2013)[48] conducted a study involving a sample of 28 patients in which 14 patients were treated with 3D titanium miniplates and remaining 14 patients were treated with normal 2.0mm miniplates. They compared 2.0mm miniplates with 3-D plate and proved that 3-D plates were superior due to its quadrangle geometry and good resistance to torque forces. The 3-D plates had properties like simplicity, malleability, low profile and ease of application. Both miniplate system reduced incidences of postoperative infections and need for IMF.

B.T. Suer etal (2014)[29] compared stability and resistance to mechanical forces between newly designed six hole non-compression titanium miniplate and six hole non-compression titanium straight miniplate. 15 fresh frozen cadaveric sheep hemimandible was used as 2 groups and 6 subgroups. Vertical, lateral and tensile forces were applied to the plated angle fracture which were fixed with 2.0x6.0mm monocortical screws according to Champys miniplate osteosynthesis principle. Loads were applied until the displacement of 4 mm was achieved and the loads were recorded. On comparison the new miniplate provided more biomechanical stability than the conventional Champy technique. But this new miniplate could be used only when the fracture is non-comminuted, non-complicated and minimally displaced angle fracture. Using single 2.0mm non-compression miniplate in angle fracture showed improved resistance to tortional displacing forces.

Al-Moraissi E.A. etal (2014)[52] studied and compared the clinical and radiological outcomes of three-dimensional (3D) miniplates and standard miniplates fixed in mandibular angle fractures according to Champy's principles. A clinical study was carried out on 20 patients with mandibular angle fractures divided into two groups. Group A patients were treated with a single 1-mm 3D titanium miniplate; group B patients were treated with a single 2.0-mm standard titanium miniplate. Patients were followed for 6 months for infection, wound dehiscence, segmental mobility, malocclusion, mouth opening, hardware failure, hardware palpability, paraesthesia, and malunion/non-union. A densitometry analysis was performed using DIGORA software on digital panoramic radiographs to evaluate bone healing. No major difference in terms of the radiographic assessment was observed between the two systems. Standard miniplates & 3D plates were comparatively effective treatment modality in relation to the complications that occurred during the study period.

G. P. de Jesus etal (2014)[37] did a study with finite element model on three different methods for osteosynthesis of low subcondylar fractures: (1) two four-hole straight plates, (2) one seven-hole lambda plate, and (3) one four-hole trapezoidal plate. Load was applied to the first molar on the contralateral side to the fracture and analysed for displacement and tension distribution. The three methods were capable of withstanding functional loading. The lambda plate displayed a more homogeneous stress distribution. This method was better than single miniplate fixation.

K.P. de Oliveira etal (2014)[33] comparatively studied the mechanical resistance of square and rectangular 2.0mm 3D miniplates with standard 2 straight miniplates in the simulated fracture site of symphysis & parasymphysis of mandible. Six groups of polyurethane mandibles containing 90 standardized replicas were compared. 60 straight 4 hole miniplates used as standard pattern, 30 four hole square miniplates and 30 four hole rectangular miniplates fixed with 6mm and 12 mm long 2.0mm miniscrews. Vertical load was applied with the help of Universal Testing Machine set to a velocity of 10mm/min to all groups at left first molar. The greatest displacement was registered on the rectangular plates when compared to square and straight miniplates in the symphyseal and parasymphyseal region.

Stefano Benazzi etal (2014)[36] studied masticatory loadings which produced high stresses that were transmitted from the occlusal surface through the body of the tooth to the dental supporting structure. This reflected a response to cope with biomechanical and functional demands of mastication using 3D finite element analysis. Using Occlusal Finger print Analyser (OFA) the occlusal contact areas were identified. Perpendicular, tangential & sliding forces were applied to the premolar. High Tensile stresses were observed in the mesiobuccal wall extending to the distobuccal wall when a force of 100N was applied. This study showed that the premolar experiences high tensile stresses during masticatory load.

Trivellato P. F. B. etal (2014)[41] evaluated the mechanical resistance of 2.0mm titanium miniplate applied to the mandibular angle either with or without continuity of the inferior border of the mandible in vitro. It was divided into two groups: favourable and unfavourable to treatment. Constant load applied at three different points. Load values were obtained at three different moments of displacement: 1 mm, 2 mm and fixation failure. When the load was applied at the central incisors, a statistically significant difference was observed at fixation failure in the unfavourable to treatment group, with better results in the subgroup with continuity of the inferior border of the mandible compared to the subgroup without continuity. Discontinuity of the inferior border of the mandible did not decrease the mechanical resistance of the fixation.

Yang L. & Patil P.M. (2015)[46] did a prospective randomized study to evaluate the efficacy of a 2.0mm locking plate and screw system compared with a 2.0-mm non-locking plate and screw system in fixation of 60 patients with isolated non-comminuted mandibular angle fractures and followed up for 6 months. Based on the complication rates determined statistically conventional plates and screw system had less stability and resistance when compared to locking plates and screw system.

MATERIALS & METHODS

METHODOLOGY:

i. Selection of a human cadaveric mandible - dentulous with the presence of third molars.

ii. CT scan of the mandible was done with 3D reconstruction using DICOM software.

iii. All the miniplates and screws were scanned under ATMOS Compact Scan 2M (GOM Technology) – 3D Scanner before and after bending using their manufacturer specified plate benders.

iv. All the plates were measured for dimensions and profile changes prior to and after bending and twisting using digital Vernier Calliper and Precision Vision Measuring System (VMS) RAPID-I.

v. Initially a 3D model of mandible with simulated fractures at the symphysis region and angle region was created, 2.0mm Conventional & Locking Titanium straight 4-hole with gap miniplates & screws post bending were created and all the plates were adapted in place of the fracture site and 8 individual fracture simulated models were created using part modelling software i.e., CATIA V5.

vi. Components and mandible were saved in IGES format.

vii. The same model was imported to the HyperMesh, pre-processor software, for meshing and applying boundary and load conditions.

viii. The model was then exported to Ansys software and analysis was performed. Results were analysed in the same software.

MATERIALS AND METHODS:

TABLE 1: MATERIALS

S.No	MATERIAL	BRAND, MANUFACTURER
1.	Cadaveric Dentulous human mandible (all teeth erupted including third molar)	
2.	2.0mm Conventional Titanium straight miniplates – 4 hole with gap & 2.0 x 8mm conventional screws	Orthomax, BeMedica Health Care, India.
3.	2.0mm Conventional Titanium straight miniplates – 4 hole with gap & 2.0 x 8mm conventional screws	Leforte system, JEIL medical corporation, Korea.
4.	2.0mm Locking Titanium straight miniplates – 4 hole with gap & 2.0 x 8mm Locking screws	Leforte system, JEIL medical corporation, Korea.
5.	2.0mm Locking Titanium straight miniplates – 4 hole with gap & 2.0 x 8mm Locking screws	Synthes, Johnson & Johnson, North America.

TABLE 2: EQUIPMENTS

S.No	PROCEDURE	INSTRUMENT/SOFTWARE
1.	CT scan of the mandible	SIEMENS SOMATOM Definition Flash, 256 slice dual source stellar detector, Germany
2.	Conversion of CT image to computerized model	MIMICS 14.0
3.	3D scanning and modelling of titanium miniplates and screws	ATMOS Compact Scan 2M (GOM Technology) – 3D Scanner, Germany
4.	To Bend the miniplates	Plate benders of the specificed companies
5.	To measure length & width of plates and screws	Electronic/digital Vernier Caliper, Germany.
6.	To measure hole dimensions and profile inspection of miniplates before and after manipulation	Precision Vision Measuring System (VMS), RAPID-I, Model - V2015 LX and V4020 LX, Customised Technologies (P) Ltd., India.
7.	To measure human bite forces	AXPERT bite force measuring device, India.

TABLE 3: SOFTWARE

S.No	PROCEDURE	SOFTWARE	MANUFACTURER
1.	CAD modelling	CATIA V5	Dassault systemes, Velizy Villacoublay, France.
2.	Finite element modelling	Altair hypermesh	Altair engineering Inc., Michigan, USA.
3.	Meshing	Shell 181	Ansys Inc., Pennsylvania, USA.
4.	Analysis	Ansys 14.5	Ansys Inc., Pennsylvania, USA.

SCHEMA OF THE STUDY

Selection of a human cadaveric mandible dentulous with the presence of third molars.

↓

CT scan of the mandible was done with 3D reconstruction using MIMICS software.

↓

3D Scanning before and after bending of miniplates using their manufacturer specified plate benders.

↓

Dimensions and profile changes prior to and after bending and twisting using digital Vernier Calliper and Precision Vision Measuring System(VMS), RAPID-I

↓

8 individual fracture models fixed with miniplates were created using Part Modelling software – CATIA V5

↓

Finite Element Modelling in Hypermesh

↓

Meshing in SHELL 181

↓

Analysis of load and boundary conditions

↓

Analysis in Ansys 14.5 software

↓

Validation of results

1) Selection of Cadaveric human mandible and 3D reconstruction of the model with CT data.

Cadaveric human mandible with all the teeth erupted including the third molars in the arch was taken for the study and CT scanning of the mandible done. 3D reconstruction of the mandible was achieved using MIMICS software. Ethical clearance was obtained.

2) Miniplates 3D scanning and measurements

2.0mm conventional and locking titanium miniplates and conventional and locking head screws of different designs were measured prior to and after bending using digital Vernier Caliper and VMS – RAPID-I. All the plates were bent by single operator for standardization. Four miniplates for each plate designs were taken into consideration and two plates for symphysis region and one for the angle region was bent and adapted. After which the miniplates were placed under 3D ATMOS scanner and all the components were scanned for adaptation to the fracture simulated mandibular 3D model.

3) CAD Modelling

Using a CAD program the mandible and miniplates were modelled by using CATIA (Computer Aided Three Dimensional Interactive Applications) V5 software. CATIA V5 products are organized on three different platforms like P1, P2, and P3. In this study, platform P3 was used to do the modelling.

- **Steps for creating the model**

i. Initially a 2D sketch of the mandible was created using profile toolbar which is used to create simple geometries.

ii. Dimensions and constraints are given to the sketch to restrict its degrees of freedom and make it stable.

iii. The 2D sketch was then converted into 3D model by using an option called rib.

4) Finite element model

CAD model was then imported to a meshing software tool to generate mesh. The final meshed model consists of several elements that collectively represent the entire structure. The elements not only represent segments of the structure, they also simulate it's mechanical behavior and properties. Meshing of the Mandibular structure is done in Altair HyperMesh V11.

Material properties

Material properties were derived by using formulas to perform modelling. The material properties used in the previous studies were similar to the materials we selected for this study. These values are incorporated into the study model. The values are tabulated below (Table 4).

Table 4: Material properties

	Young's modulus (GPa)	Poisson's ratio
Enamel	83.53	0.3
Dentin	17.67	0.31
Pulp	0.002	0.45
Periodontal Ligament	0.0689	0.45
Human cortical bone	13.7	0.3
Human cancellous bone	1.37	0.3
Titanium miniplates & screws	11.0	0.34
Coefficient of sliding friction for Temporomandibular joint		0.334

5) Application of boundary conditions and pressure

Applying boundary conditions: It is necessary to apply boundary conditions to simulate the different movements of the FE model while carrying out the analysis. The single mandibular FE model was used for all the 8 plating model conditions. The type of analysis is static i.e., analysis of structure subjected to load when the mandible is at rest.

Applying pressure: The total force acting was applied on the occlusal surface of the dentulous mandible.

6) Force Values for different clinical conditions

As 2 different clinical conditions (symphysis & angle fracture) are considered in this study, different areas where forces applied are:

1) For right angle fracture forces are applied over right molars, right premolars & right anteriors.

2) For symphysis fracture forces applied over bilateral molars and premolars.

This is used to evaluate the Von Mises stress over the plates and the adjacent bone over FEA model. The forces of occlusion to be evaluated were determined from various sources of literature.

For right angle fracture the left side was considered as the non-working side and the right side was considered as the working side and the load was applied to mimic the final point of closure in lateral masticatory movement. Different magnitude of forces was applied for each area of the dentition which were obtained as an average between control group and patients who got operated for symphysis and angle fracture. Ethical clearance was obtained and patient's consent was obtained to measure the bite forces using AXPERT bite force gauge.

TABLE 5: Statistical Average Bite forces obtained from 2months post operated group

TOOTH	VALUE (Kg) (Average)	VALUE (N) (Average)
RIGHT MOLAR	22.65273	226.5293
RIGHT PREMOLAR	19.76127	197.6127
ANTERIORS	13.72393	137.2393
LEFT PREMOLAR	22.45033	224.5033
LEFT MOLAR	25.61767	256.1767

The limitations for these bite forces was pain and was measured using numeric pain rating scale which was given to the patient.

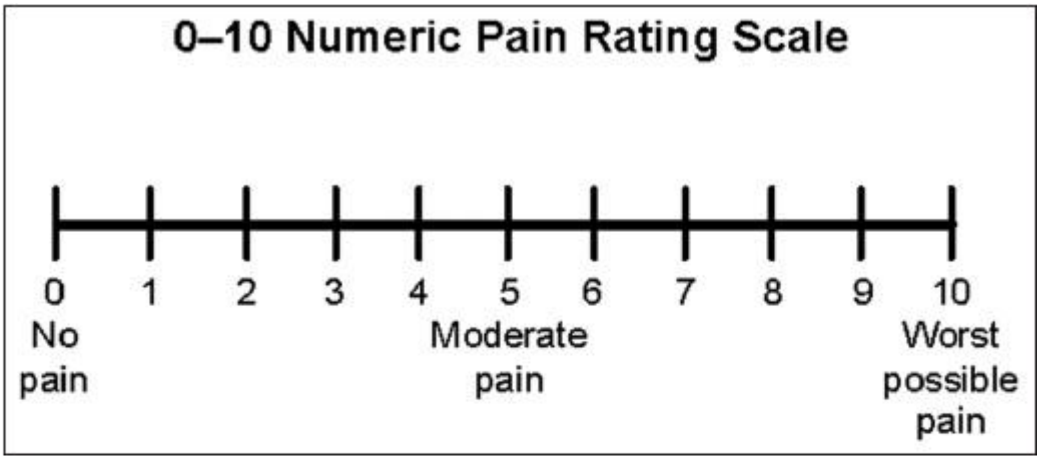


TABLE 6: NUMERIC PAIN RATING SCALE – PRE OPERATIVE

S.No	PATIENT NAME	AGE	SEX	0	1	2	3	4	5	6	7	8	9	10
1	JAGADEESH	26	M											X
2	VELUSAMY	50	M										X	
3	GOWTHAMAN	22	M											X
4	KANAGARAJ	30	M											X
5	KAMESH	21	M											X
6	SUNDAR	30	M											X
7	SAMPATH KUMAR	26	M											X
8	PALANISAMY	26	M											X
9	RAVI	33	M									X		
10	RAGHUVEER	24	M											X
11	SENTHIL KUMAR	46	M											X
12	SANTOSH	28	M											X
13	SATISH KUMAR	24	M											X
14	DAVID	19	M											X
15	SAKTHIVEL	25	M											X
16	VIJAYARAGHAVAN	34	M											X
17	DHANASEKAR	20	M											X
18	MURUGAN	48	M											X
19	AVINASH	23	M										X	
20	RAVICHANDRAN	43	M									X		

X – PAIN SUFFERED BY THE PATIENT

--- - PATIENT DID NOT VISIT BACK SO EXCLUDED FROM THE STUDY

TABLE 7: NUMERIC PAIN RATING SCALE – 1 WEEK POST OPERATIVE

S.No	PATIENT NAME	AGE	SEX	0	1	2	3	4	5	6	7	8	9	10
1	JAGADEESH	26	M						X					
2	VELUSAMY	50	M							X				
3	GOWTHAMAN	22	M							X				
4	KANAGARAJ	30	M								X			
5	KAMESH	21	M							X				
6	SUNDAR	30	M						X					
7	SAMPATH KUMAR	26	M							X				
8	PALANISAMY	26	M								X			
9	RAVI	33	M								X			
10	RAGHUVEER	24	M						X					
11	SENTHIL KUMAR	46	M							X				
12	SANTOSH	28	M							X				
13	SATISH KUMAR	24	M								X			
14	DAVID	19	M								X			
15	SAKTHIVEL	25	M						X					
16	VIJAYARAGHAVAN	34	M							X				
17	DHANASEKAR	20	M								X			
18	MURUGAN	48	M							X				
19	AVINASH	23	M						X					
20	RAVICHANDRAN	43	M					X						

X – PAIN SUFFERED BY THE PATIENT

--- - PATIENT DID NOT VISIT BACK SO EXCLUDED FROM THE STUDY

TABLE 8: NUMERIC PAIN RATING SCALE – 1 MONTH POST OPERATIVE

S.No	PATIENT NAME	AGE	SEX	0	1	2	3	4	5	6	7	8	9	10
1	JAGADEESH	26	M			X								
2	VELUSAMY	50	M				X							
3	GOWTHAMAN	22	M					X						
4	KANAGARAJ	30	M				X							
5	KAMESH	21	M				X							
6	SUNDAR	30	M				X							
7	SAMPATH KUMAR	26	M				X							
8	PALANISAMY	26	M			X								
9	RAVI	33	M			X								
10	RAGHUVEER	24	M	X										
11	SENTHIL KUMAR	46	M	-	-	-	-	-	-	-	-	-	-	-
12	SANTOSH	28	M		X									
13	SATISH KUMAR	24	M			X								
14	DAVID	19	M			X								
15	SAKTHIVEL	25	M		X									
16	VIJAYARAGHAVAN	34	M		X									
17	DHANASEKAR	20	M	X										
18	MURUGAN	48	M		X									
19	AVINASH	23	M			X								
20	RAVICHANDRAN	43	M		X									

X – PAIN SUFFERED BY THE PATIENT

--- - PATIENT DID NOT VISIT BACK SO EXCLUDED FROM THE STUDY

TABLE 9: NUMERIC PAIN RATING SCALE – 2 MONTHS POST OPERATIVE

S.No	PATIENT NAME	AGE	SEX	0	1	2	3	4	5	6	7	8	9	10
1	JAGADEESH	26	M	X										
2	VELUSAMY	50	M	X										
3	GOWTHAMAN	22	M	X										
4	KANAGARAJ	30	M	X										
5	KAMESH	21	M		X									
6	SUNDAR	30	M	X										
7	SAMPATH KUMAR	26	M	X										
8	PALANISAMY	26	M	X										
9	RAVI	33	M	X										
10	RAGHUVEER	24	M	X										
11	SENTHIL KUMAR	46	M	-	-	-	-	-	-	-	-	-	-	-
12	SANTOSH	28	M	X										
13	SATISH KUMAR	24	M	X										
14	DAVID	19	M		X									
15	SAKTHIVEL	25	M	X										
16	VIJAYARAGHAVAN	34	M	X										
17	DHANASEKAR	20	M	-	-	-	-	-	-	-	-	-	-	-
18	MURUGAN	48	M	-	-	-	-	-	-	-	-	-	-	-
19	AVINASH	23	M	-	-	-	-	-	-	-	-	-	-	-
20	RAVICHANDRAN	43	M	-	-	-	-	-	-	-	-	-	-	-

X – PAIN SUFFERED BY THE PATIENT

--- - PATIENT DID NOT VISIT BACK SO EXCLUDED FROM THE STUDY

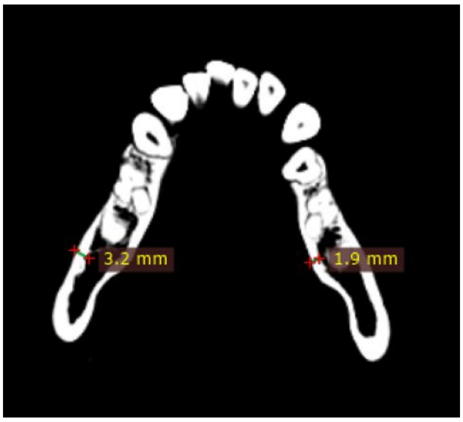
7) Analysis

In this step the finite element model was exported to the solver software to carry out analysis. Both solving and post processing of fracture simulated mandibular model was done in Ansys V14.5.

- **Post processing**

This is the final step in a finite element analysis. Results obtained in the previous phase are usually in the form of raw data and difficult to interpret. In post analysis, a CAD program was utilized to manipulate the data for generating deflected shape of the structure, creating stress plots, animation, etc. A graphical representation of the results is very useful in understanding the behaviour of a structure. The results obtained in this phase are Von Mises stress and deformation plot.

Fig 1: Cadaveric dentulous human mandible with all teeth including third molar present



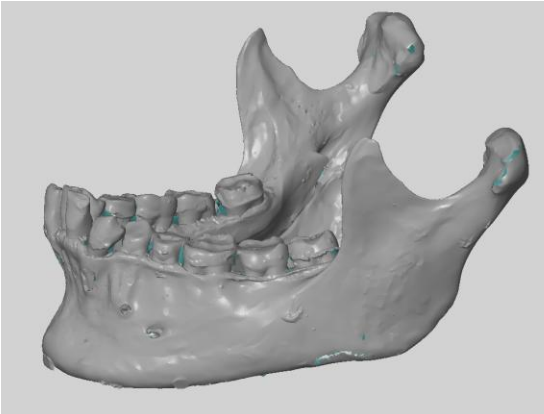


Fig 2, 3: (*from left to right*) CT scan of the mandible converted into 3D model

4A

4B

4C

4D

Fig 4A – Synthes Locking plate, 4B – Orthomax conventional plate, 4C – Leforte conventional plate, 4D – Leforte Locking plate

Fig 5, 6: (*from left to right*) Miniplate thickness measured using digital Vernier Caliper

Fig 7: Rapid – I – Vision Measuring System with miniplate

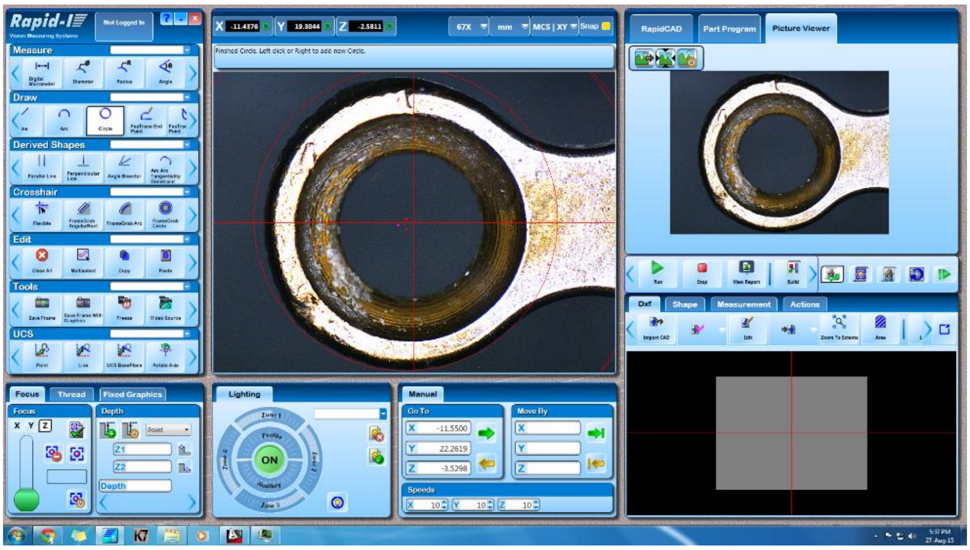


Fig 8: Screen shot of the measurement progress of the miniplate

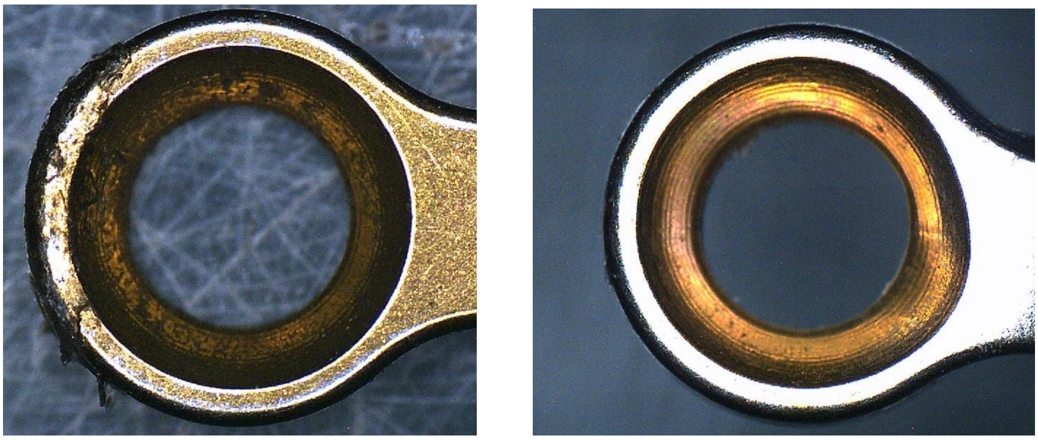


Fig 9, 10: (*from left to right*) Original plate before bending and dimensional change shown after bending

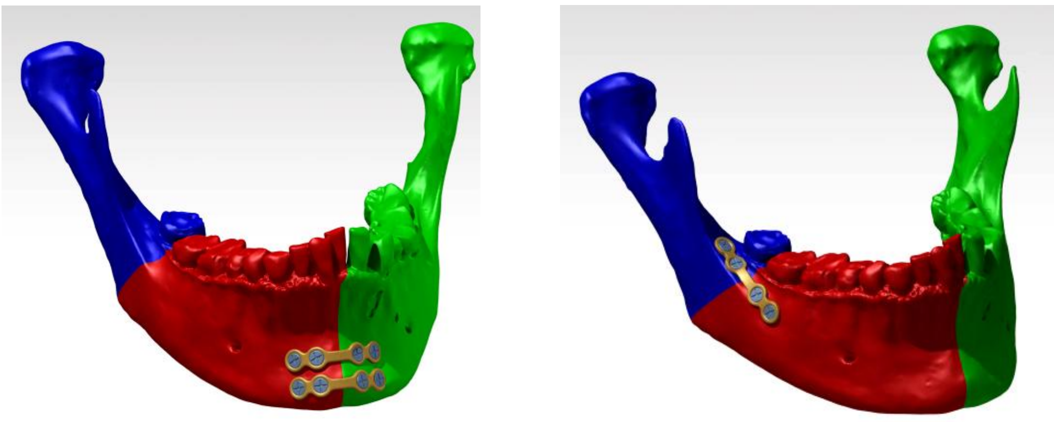
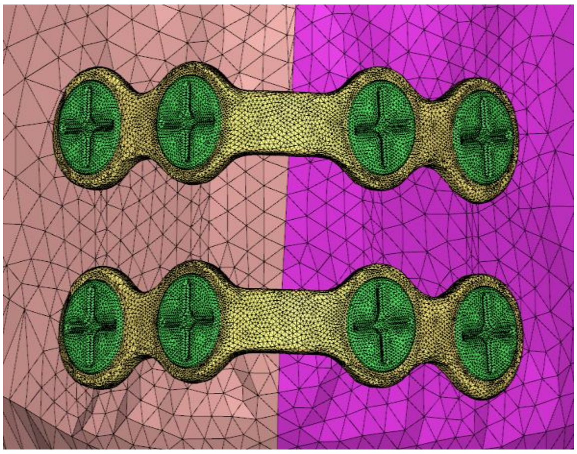


Fig 11, 12: (*from left to right*) Assembled miniplate and mandible model with simulated symphysis and angle fracture depicted in different colours



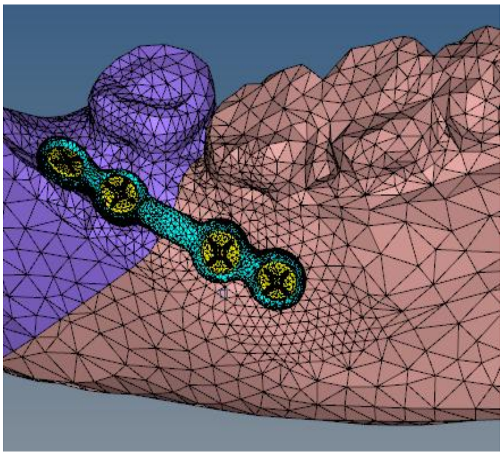


Fig 13, 14: (*from left to right*) Meshed Titanium miniplate and mandible model with simulated fracture depicted in different colours

Fig 15: AXPERT Bite Force gauge used for the study

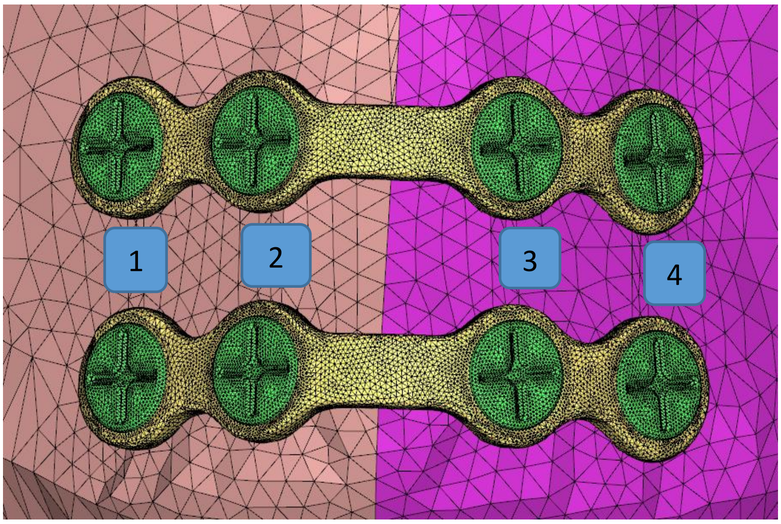


Figure 16: From right to left number of the hole marked and the dimensional change is checked under RAPID – I, both for tension and compression border plates in symphysis fracture.

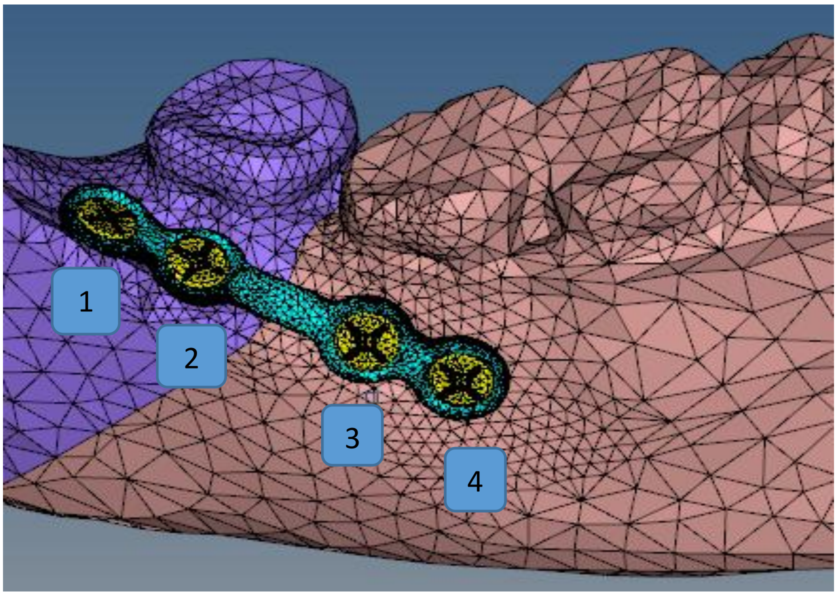


Figure 17: From top to bottom number of the hole marked and the dimensional change is checked under RAPID – I for angle fracture.

RAPID – I – VMS MEASUREMENTS

NORMAL PROFILE AND MEASUREMENTS - ORTHOMAX CONVENTIONAL PLATES

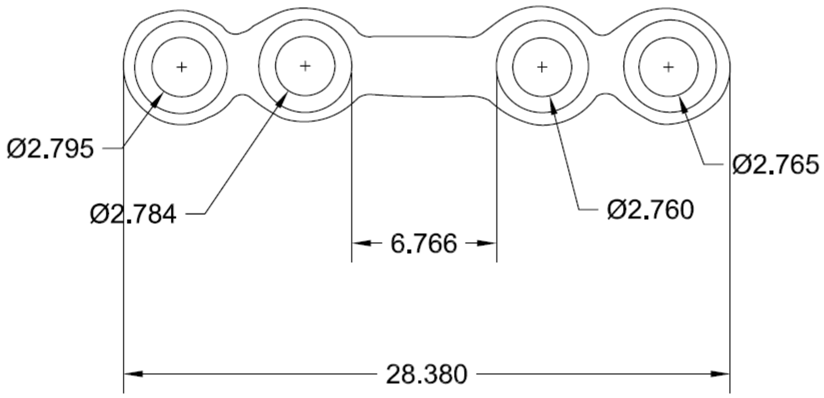


Figure 18: Profile of Orthomax Conventional Plate

ORTHOMAX CONVENTIONAL PLATES - SYMPHYSIS REGION:

TENSION BORDER

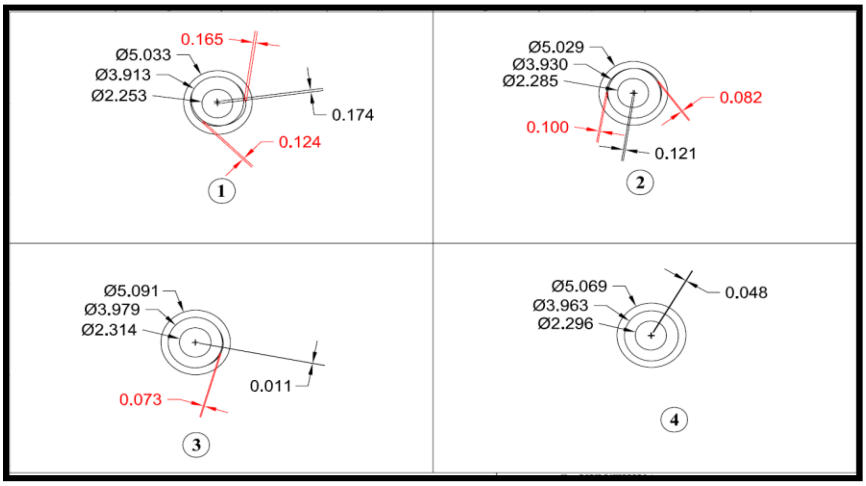


Figure 19: Tension Border Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

COMPRESSION BORDER

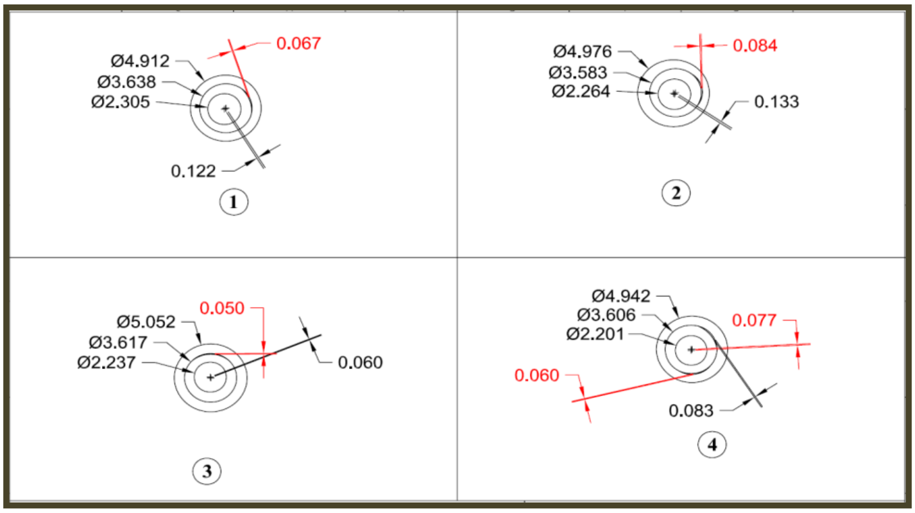


Figure 20: Compression Border Measurements

ORTHOMAX CONVENTIONAL PLATES – ANGLE REGION:

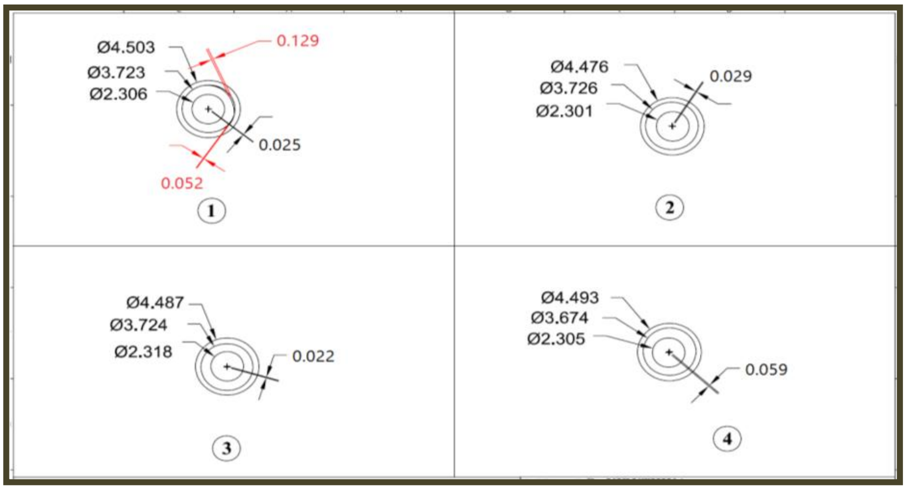


Figure 21: Angle Region Plate Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
\+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

NORMAL PROFILE AND MEASUREMENTS - LEFORTE CONVENTIONAL PLATE

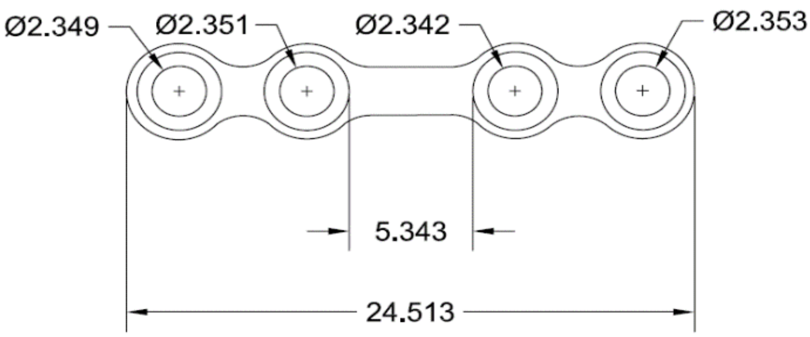


Figure 22: Profile of Leforte Conventional Plate

LEFORTE CONVENTIONAL PLATES – SYMPHYSIS REGION:

TENSION BORDER

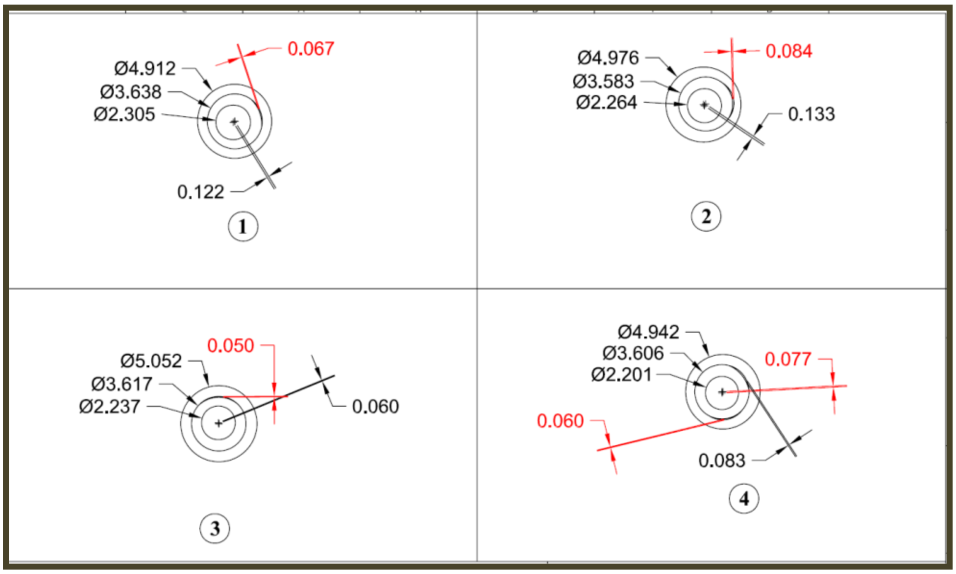


Figure 23: Tension Border Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

COMPRESSION BORDER

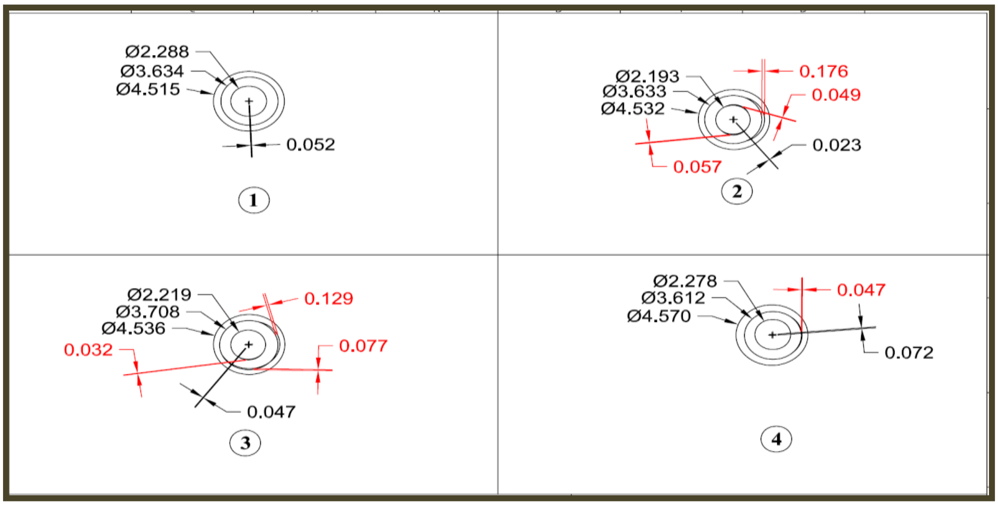


Figure 24: Compression Border Measurements

LEFORTE CONVENTIONAL PLATES – ANGLE REGION:

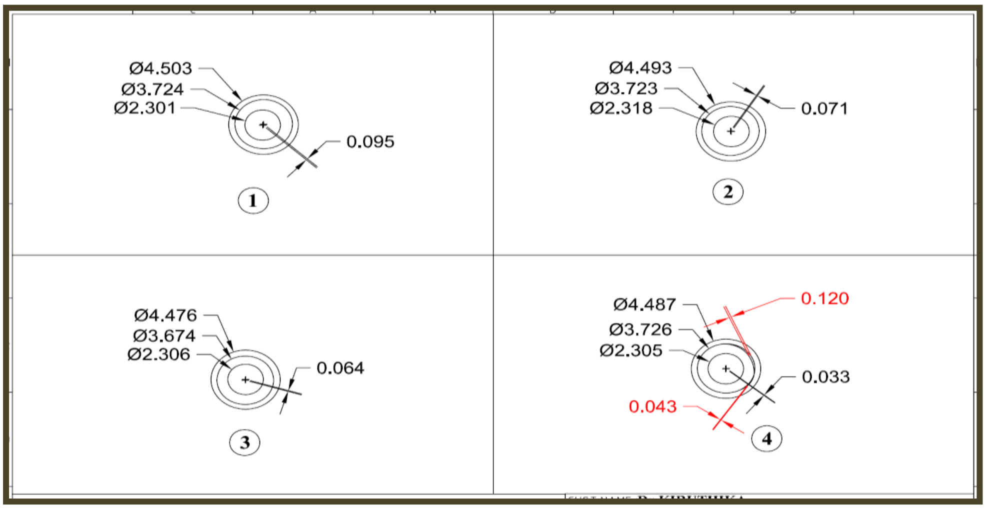


Figure 25: Angle Region Plate Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
\+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

NORMAL PROFILE AND MEASUREMENTS – LEFORTE LOCKING PLATES

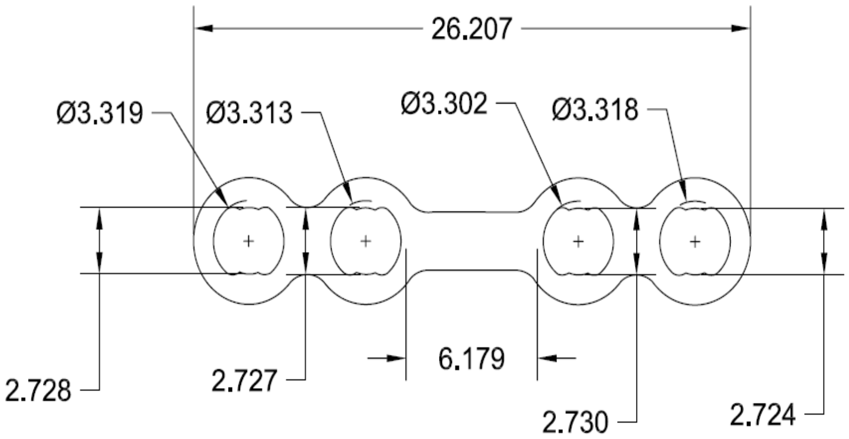


Figure 26: Profile of Leforte Locking Plate

LEFORTE LOCKING PLATES – SYMPHYSIS REGION:

TENSION BORDER

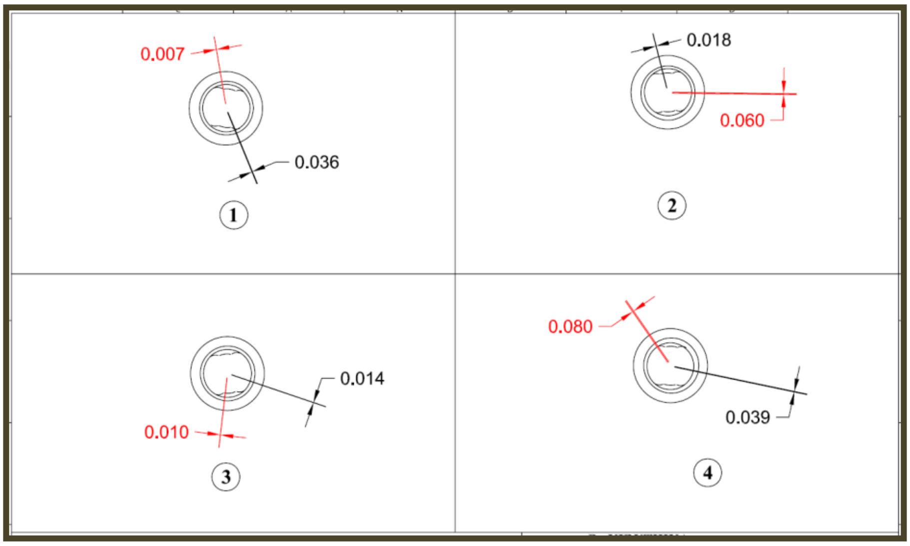


Figure 27: Tension Border Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

COMPRESSION BORDER

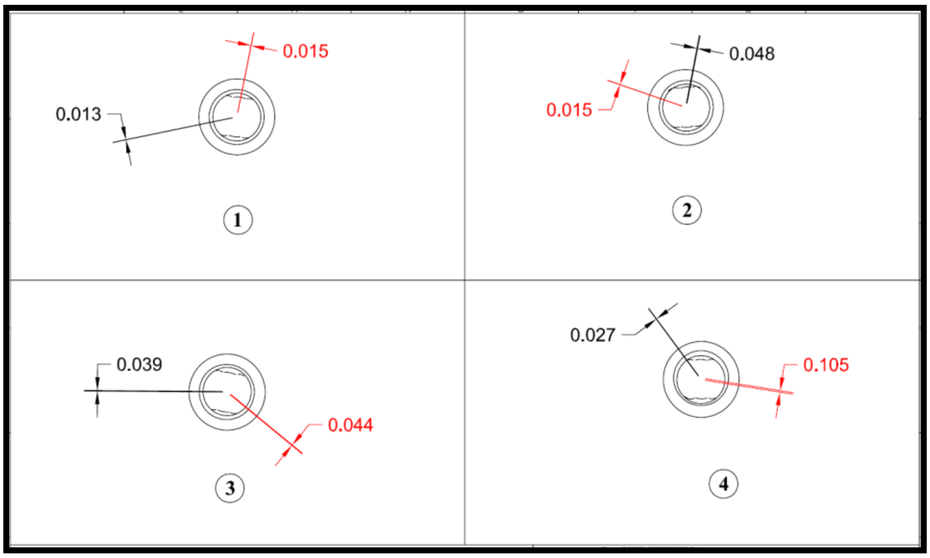


Figure 28: Compression Border Measurements

LEFORTE LOCKING PLATE – ANGLE REGION:

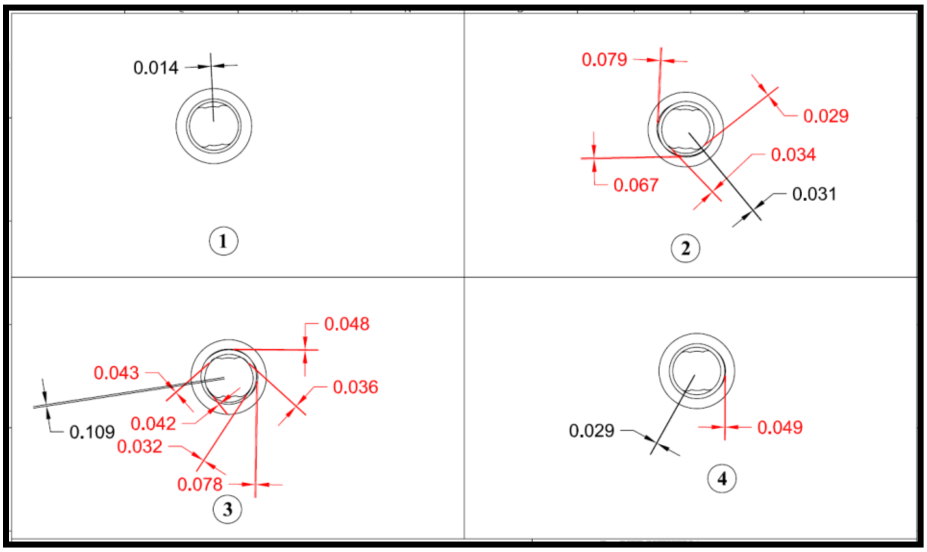
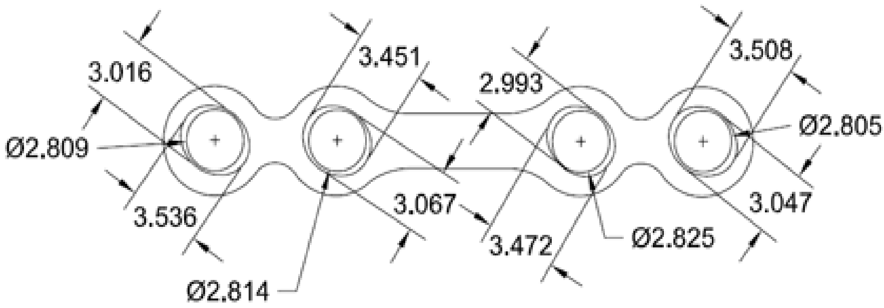


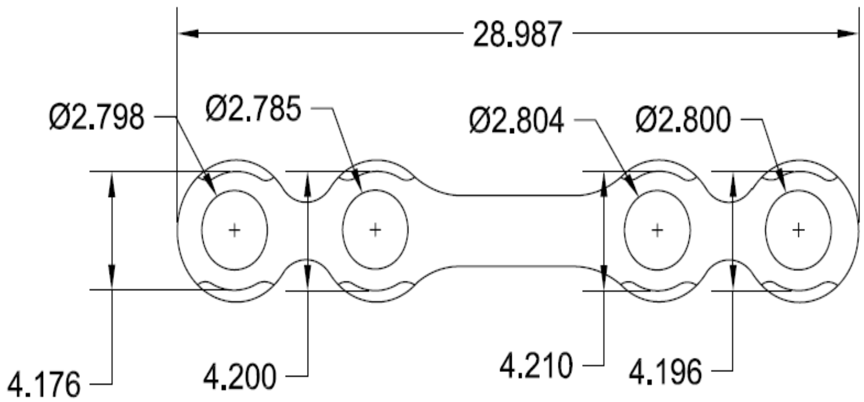
Figure 29: Angle Region Plate Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

NORMAL PROFILE AND MEASUREMENTS – SYNTHES LOCKING PLATES



FRONT SIDE



BACK SIDE

Figure 30: Profile of Synthes Locking Plate

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

SYNTHES LOCKING PLATES – SYMPHYSIS REGION:

TENSION BORDER

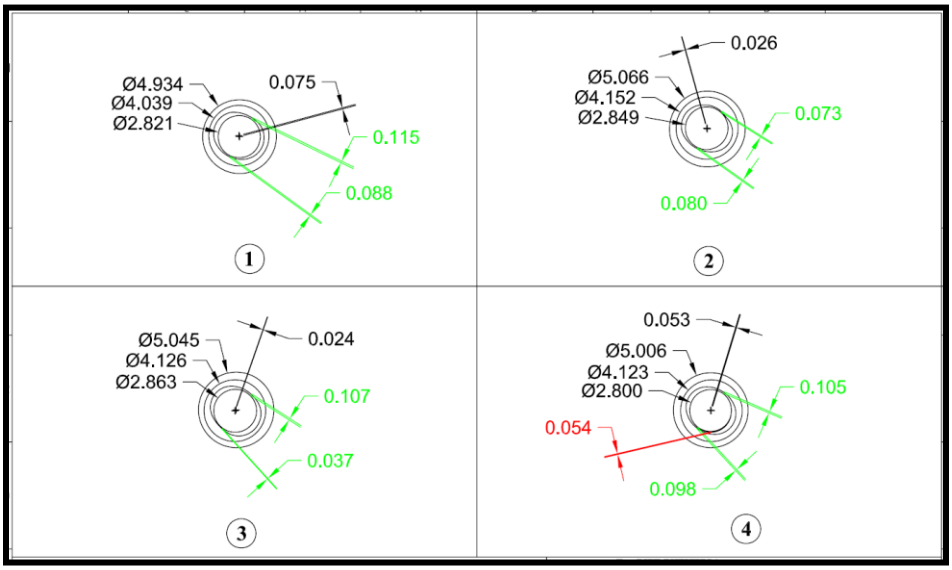


Figure 31: Tension Border Measurements

COMPRESSION BORDER

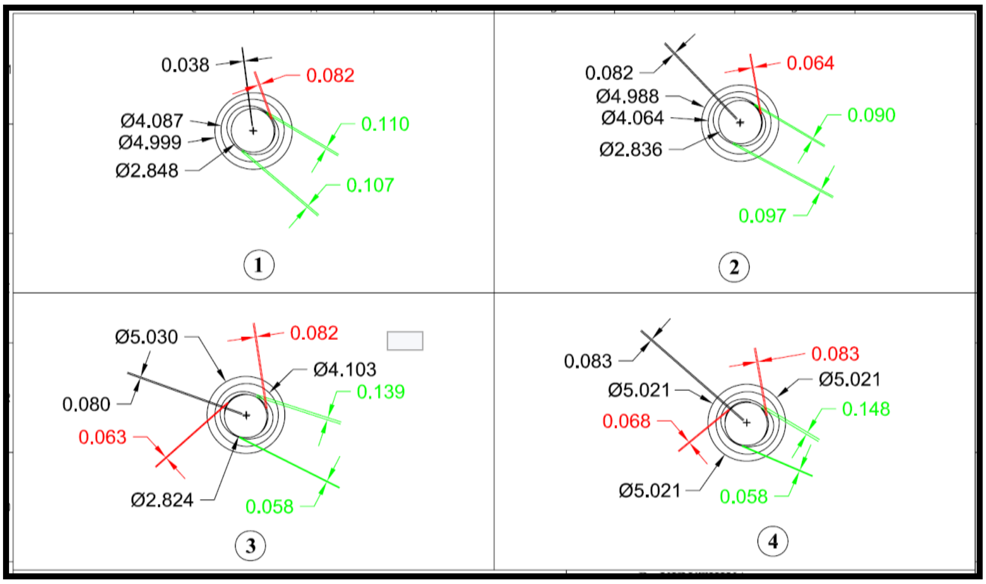


Figure 32: Compression Border Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
\+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

SYNTHES LOCKING PLATES – ANGLE REGION:

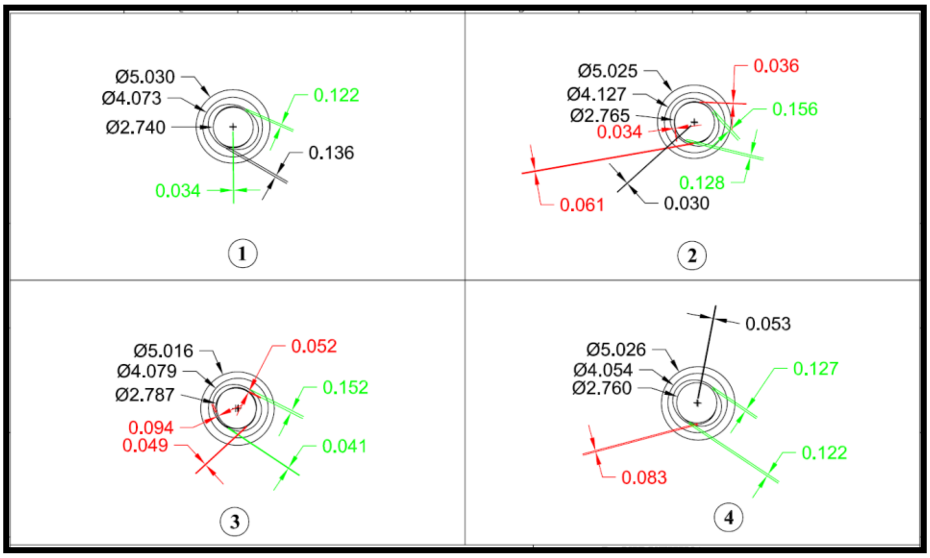


Figure 33: Angle Region Plate Measurements

NOTE: All Measurements are in Millimetres
ACCURACY → +/- 0.003mm
\+ → CENTRE OF THE CONCENTRIC CIRCLES
ϕ → DIAMETER IN MILLIMETRE
R → RADIUS IN MILLIMETRE
----- → DIFFERENCE IN DIAMETER COMPARED TO NORMAL
----- & ----- → DISTANCE BETWEEN THE ALTERED CIRCLE TO NORMAL

RESULTS

The evaluation of the Finite Element Analysis results was performed with respect to displacement and stresses in the titanium miniplates. One of the main parameters for different fracture stabilization alternatives is the fracture mobility. It is an indication of rigidity of the fixation of the 2 fractured segments of the mandible, appropriate osteogenesis and bone healing. The load across the fracture resulted in mobility of the fragments and displacement in the x, y, and z directions (Figure-34). Considering that, displacement was calculated by vectorial summarization of the movements in the x, y, and z directions. The displacement measurements for all models are presented.

The analysis was done to determine the stress over the plate & bone on vertical load application and also to determine the deformation (strain) of the bone and development of fracture gap in relation to the loading conditions. The loads were applied to simulate patients with opposing maxillary natural dentition. All stress values are given in MPa (N/mm^2).

The vertical loading forces (bite forces) cause tensile stresses in the alveolar area and compression stresses in the basal mandibular area. In the control model, the stress distribution was high in the symphysis region. With complete load, symphysis fracture gap was opened and the pressure stress got increased. With segmental load angle fracture gap was opened and the pressure stress got increased.

The segmental loading created clearly visible compression stresses in the mandibular base. After full loading, we could see stronger compression in the base. The plate generated continuity of tensile stresses in the alveolar region.

The results were analysed for the following factors:

1) Deflection & stability
2) Mechanical stress over bone
3) Maximal stress area over miniplates
4) Fracture gap and
5) Displacement direction of fracture fragment.

LOADS AND BOUNDARY CONDITIONS

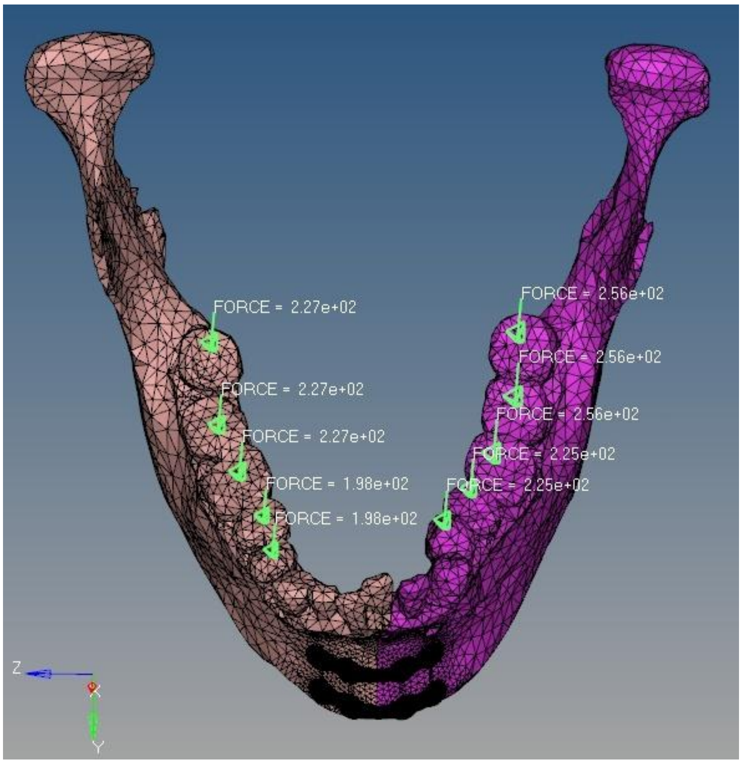
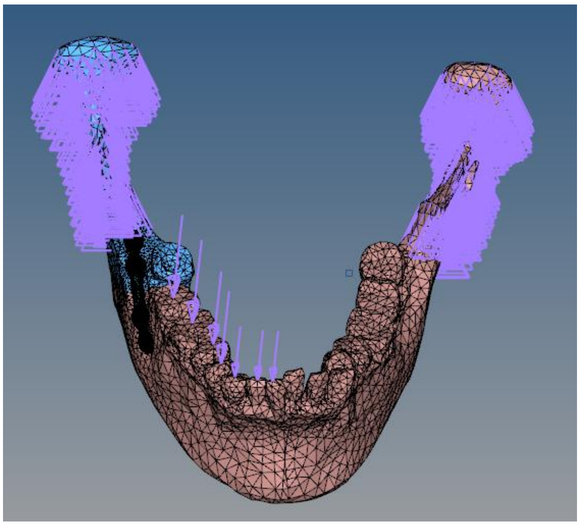


Figure 34: FORCE VALUES IN MPa (N/mm^2) APPLIED



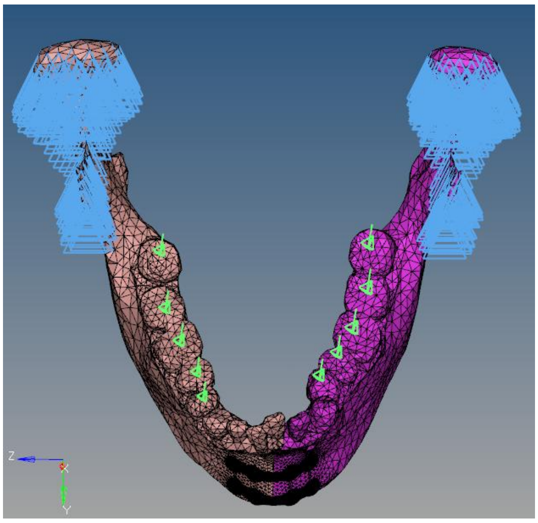


Figure 35: FORCES APPLIED FOR ANGLE FRACTURE

Figure 36: FORCES APPLIED FOR SYMPHYSIS FRACTURE

TABLE 10: MAXIMUM DEFLECTION

S.No.	FRACTURE FIXATION MODEL	MAXIMUM DEFLECTION (mm)
1	Orthomax conventional plate in symphysis region	6.05196
2	Orthomax conventional plate in angle region	5.93459
3	Leforte conventional plate in symphysis region	4.85454
4	Leforte conventional plate in angle region	2.42414
5	Leforte locking plate in symphysis region	2.50747
6	Leforte locking plate in angle region	3.45596
7	Synthes locking plate in symphysis region	6.07613
8	Synthes locking plate in angle region	3.00287

TABLE 11: MAXIMUM MECHANICAL STRESS OVER BONE

S.No.	FRACTURE FIXATION MODEL	MAXIMUM STRESS OVER BONE (MPa)
1	Orthomax conventional plate in symphysis region	98.6537
2	Orthomax conventional plate in angle region	379.81
3	Leforte conventional plate in symphysis region	92.5581
4	Leforte conventional plate in angle region	292.046
5	Leforte locking plate in symphysis region	288.611
6	Leforte locking plate in angle region	449.205
7	Synthes locking plate in symphysis region	78.476
8	Synthes locking plate in angle region	309. 63

TABLE 12: MAXIMAL STRESS OVER MINIPLATES

S.No.	FRACTURE FIXATION MODEL	MAXIMAL STRESS OVER MINIPLATES (MPa)
1	Orthomax conventional plate in symphysis region	75.4011
2	Orthomax conventional plate in angle region	2114.62
3	Leforte conventional plate in symphysis region	67.9809
4	Leforte conventional plate in angle region	1145.38
5	Leforte locking plate in symphysis region	73.3317
6	Leforte locking plate in angle region	1540.14
7	Synthes locking plate in symphysis region	61.2447
8	Synthes locking plate in angle region	833.457

TABLE 13: FRACTURE GAP

S.No.	FRACTURE FIXATION MODEL	FRACTURE GAP (mm)
1	Orthomax conventional plate in symphysis region	0.86241
2	Orthomax conventional plate in angle region	2.2708
3	Leforte conventional plate in symphysis region	0.09959
4	Leforte conventional plate in angle region	1.295088
5	Leforte locking plate in symphysis region	0.01804
6	Leforte locking plate in angle region	1.86241
7	Synthes locking plate in symphysis region	0.088748
8	Synthes locking plate in angle region	1.18295

TABLE 14: DIRECTION OF DISPLACEMENT

S.No.	FRACTURE FIXATION MODEL	DIRECTION OF DISPLACEMENT OF FRACTURED SEGMENTS
1	Orthomax conventional plate in symphysis region	Lateral
2	Orthomax conventional plate in angle region	Medial
3	Leforte conventional plate in symphysis region	Lateral
4	Leforte conventional plate in angle region	Medial
5	Leforte locking plate in symphysis region	Lateral
6	Leforte locking plate in angle region	Medial
7	Synthes locking plate in symphysis region	Lateral
8	Synthes locking plate in angle region	Medial

SYMPHYSIS FRACTURE FIXATION

SYNTHES LOCKING PLATES:

Maximum deflection on application of bilateral vertical loads is 6.07613 mm. While the Von Mises Stress over miniplates are 61.2447 MPa and over bone near the screwed area is 78.476 MPa.

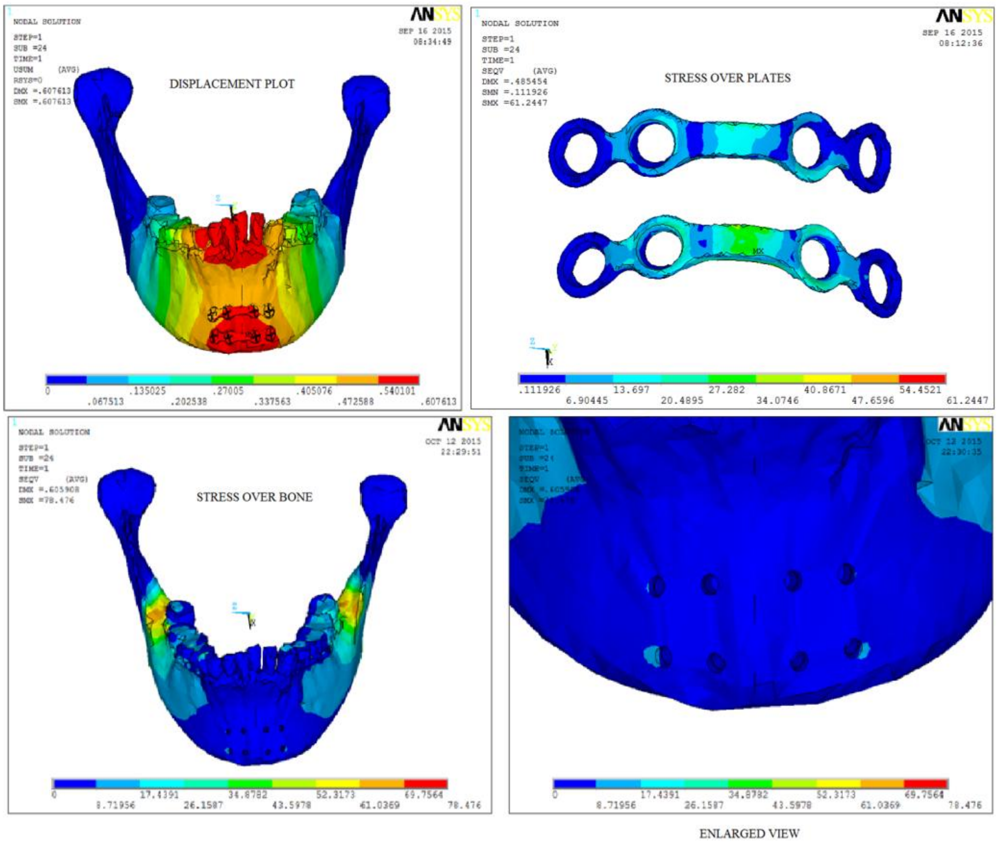


Figure 37: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

LEFORTE LOCKING PLATES:

Maximum deflection on application of bilateral vertical loads is 2.50747mm. While the Von Mises Stress over miniplates are 73.3317 MPa and over bone near the screwed area is 288.611 MPa.

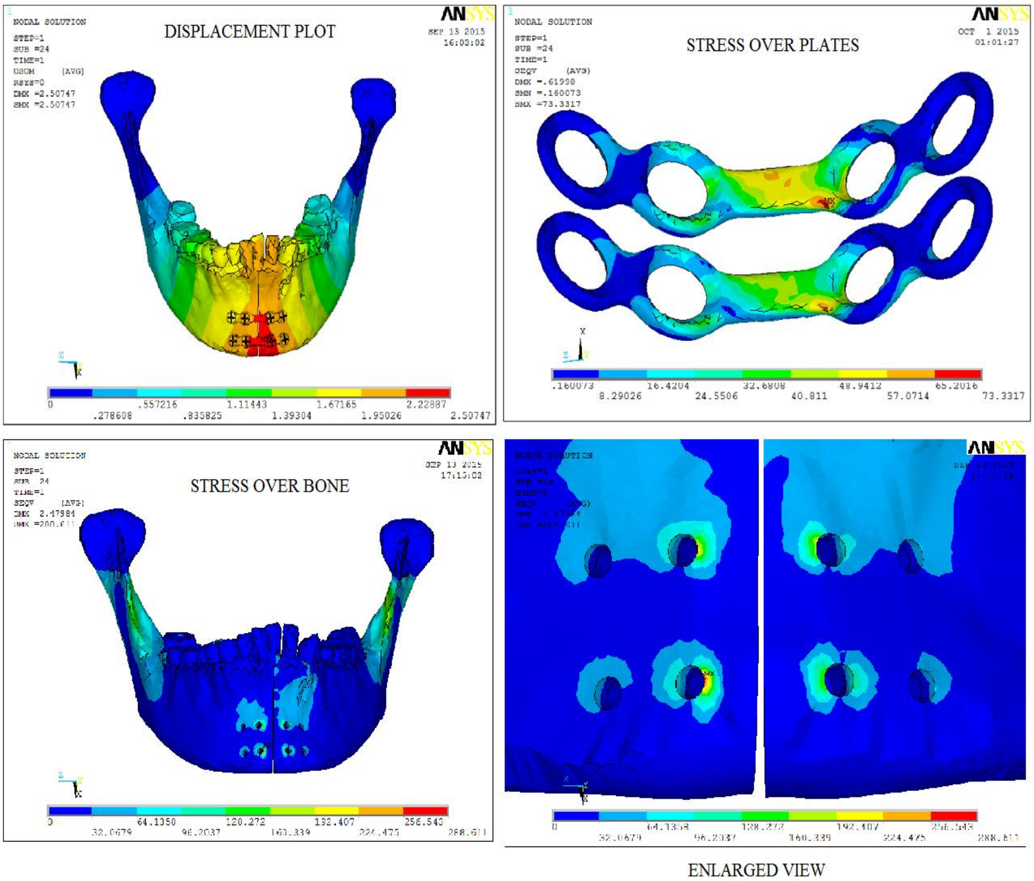


Figure 38: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

LEFORTE CONVENTIONAL PLATES:

Maximum deflection on application of bilateral vertical loads is 4.85454 mm. While the Von Mises Stress over miniplates are 67.9809 MPa and over bone near the screwed area is 92.5581 MPa.

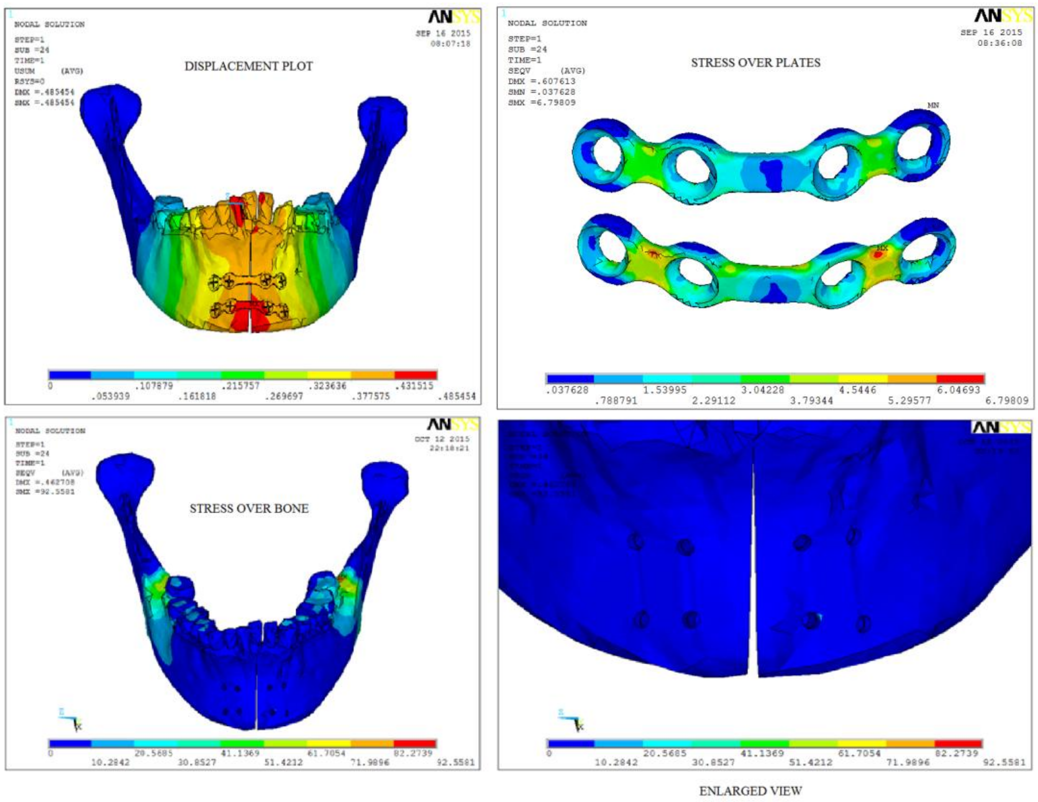


Figure 39: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

ORTHOMAX CONVENTIONAL PLATES:

Maximum deflection on application of bilateral vertical loads is 6.05196mm. While the Von Mises Stress over miniplates are 75.4011 MPa and over bone near the screwed area is 98.6537 MPa.

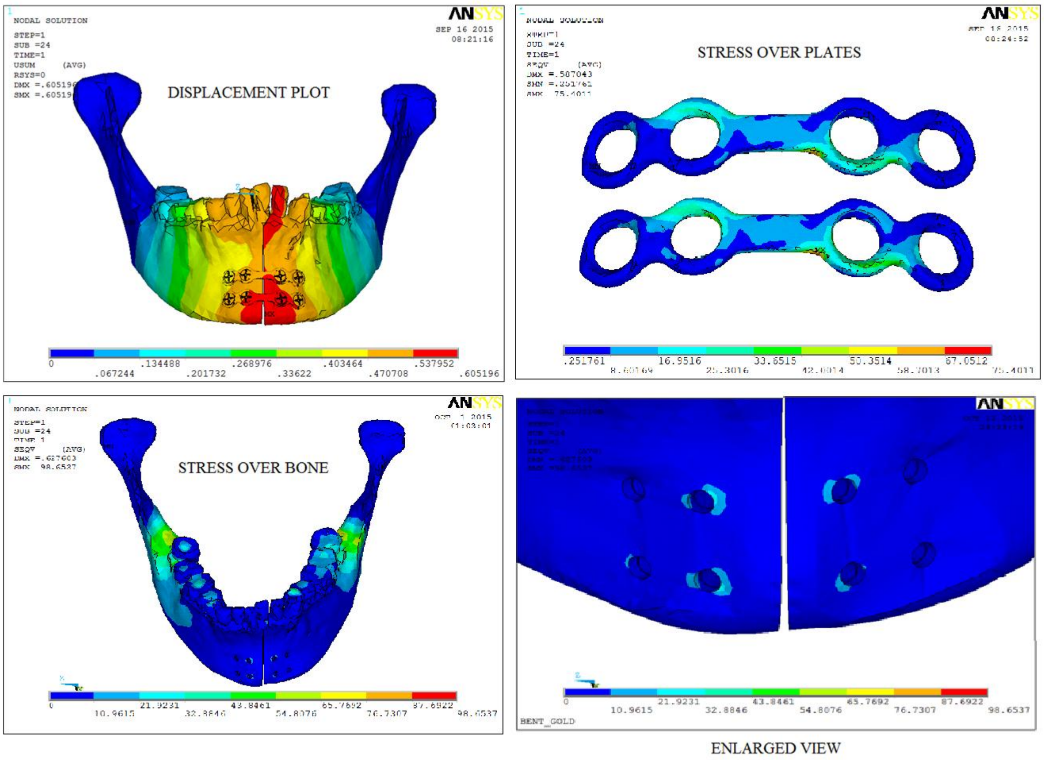


Figure 40: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

ANGLE FRACTURE FIXATION

SYNTHES LOCKING PLATES:

Maximum deflection on application of bilateral vertical loads is 3.00287 mm. While the Von Mises Stress over the miniplate is 833.457 MPa and over bone near the screwed area is 309.63 MPa.

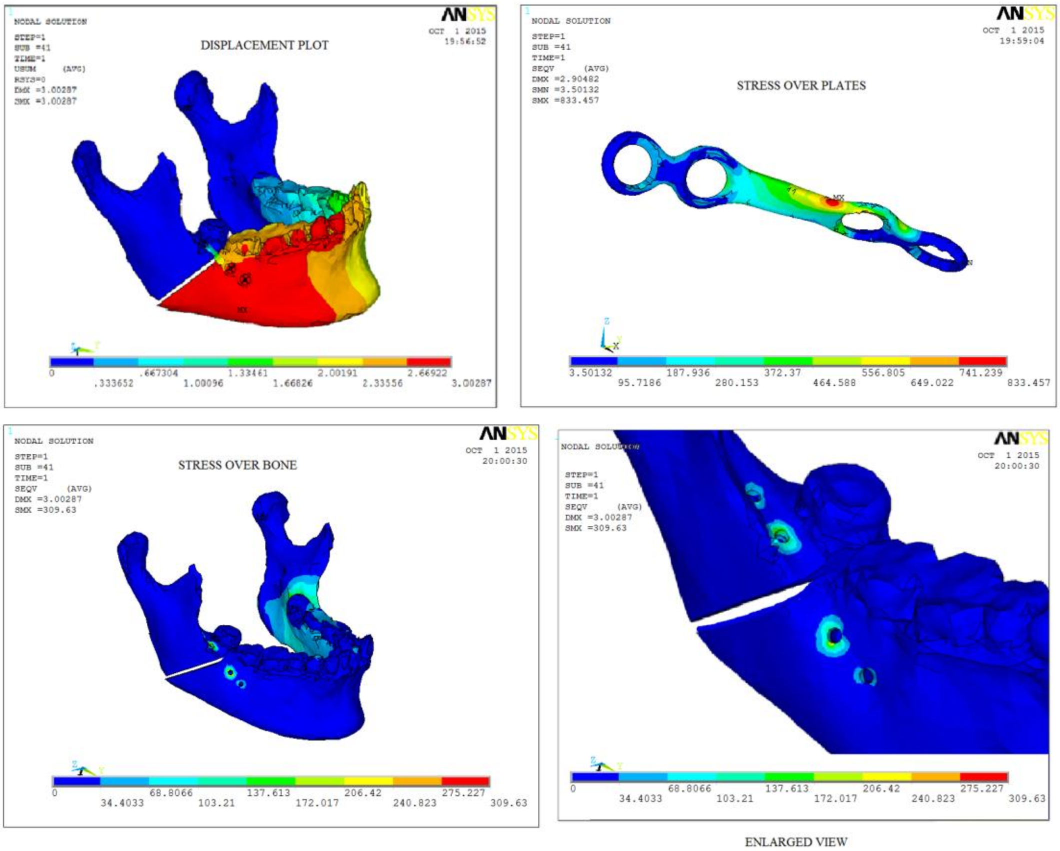


Figure 41: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

LEFORTE LOCKING PLATES:

Maximum deflection on application of bilateral vertical loads is 3.45596 mm. While the Von Mises Stress over the miniplate is 1540.14 MPa and over bone near the screwed area is 449.205 MPa.

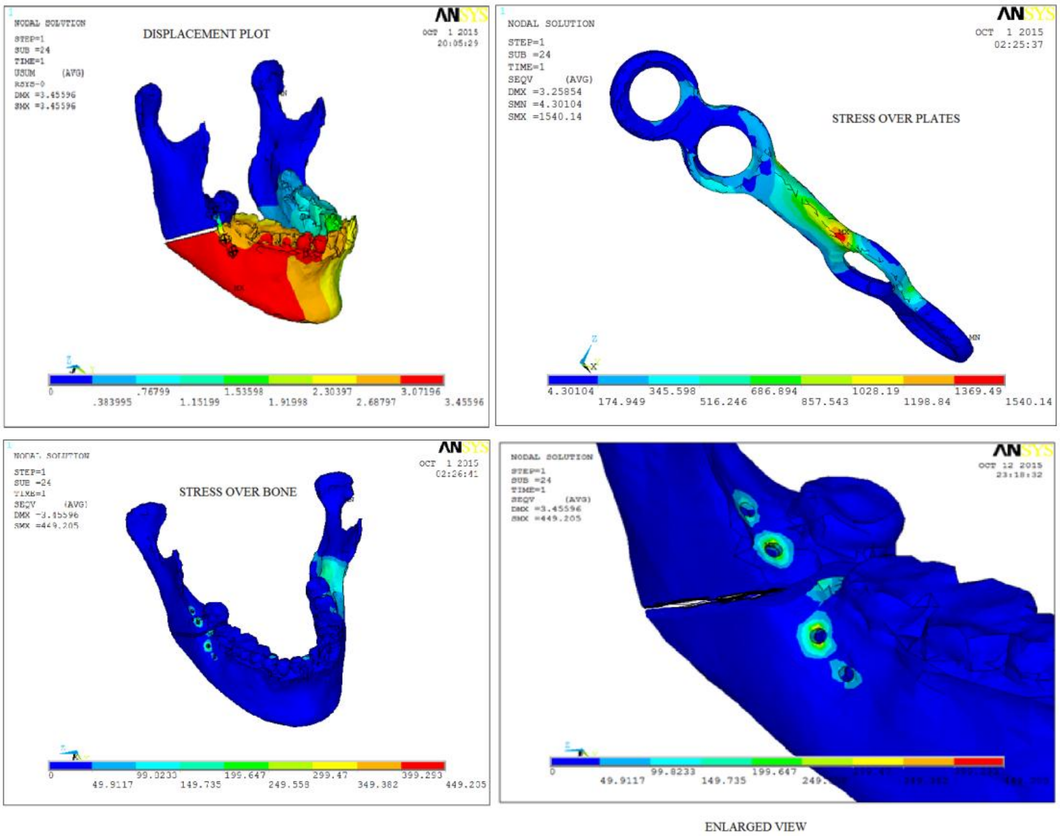


Figure 42: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

LEFORTE CONVENTIONAL PLATES:

Maximum deflection on application of bilateral vertical loads is 2.42414 mm. While the Von Mises Stress over the miniplate is 1145.38 MPa and over bone near the screwed area is 292.046 MPa.

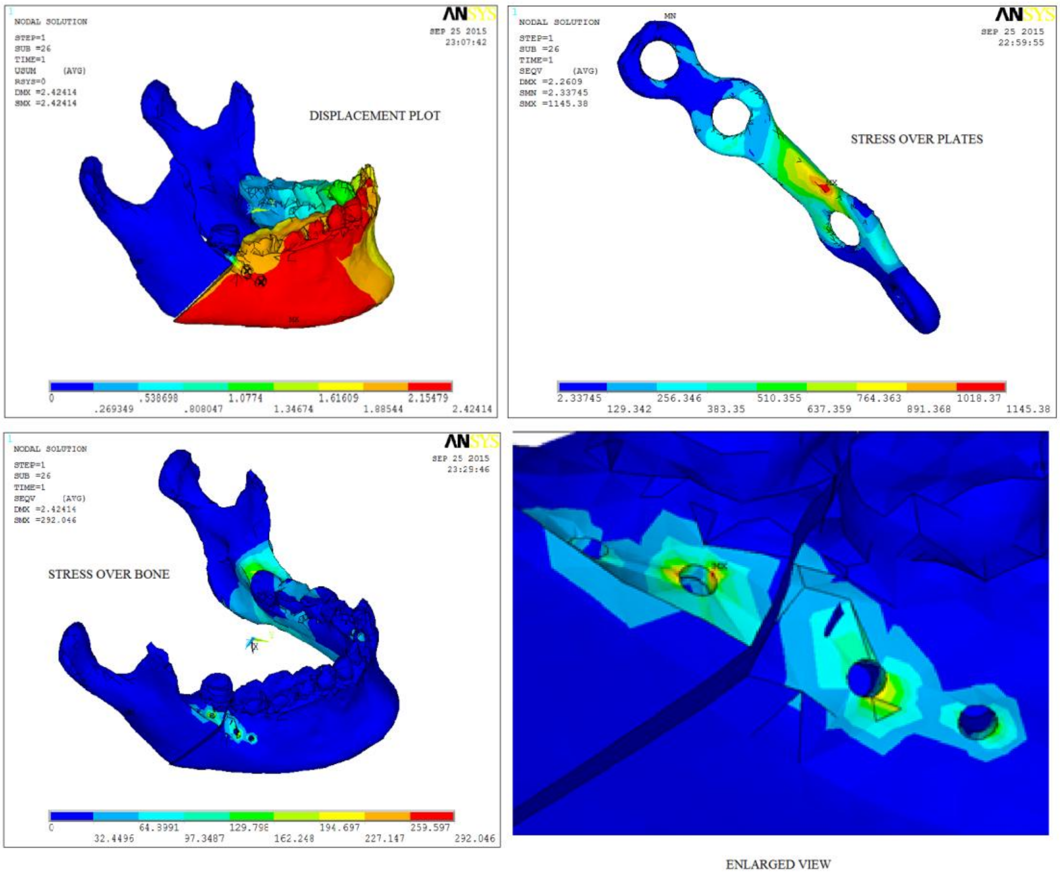


Figure 43: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

ORTHOMAX CONVENTIONAL PLATES:

Maximum deflection on application of bilateral vertical loads is 5.93459 mm. While the Von Mises Stress over the miniplate is 2114.62 MPa and over bone near the screwed area is 379.81 MPa.

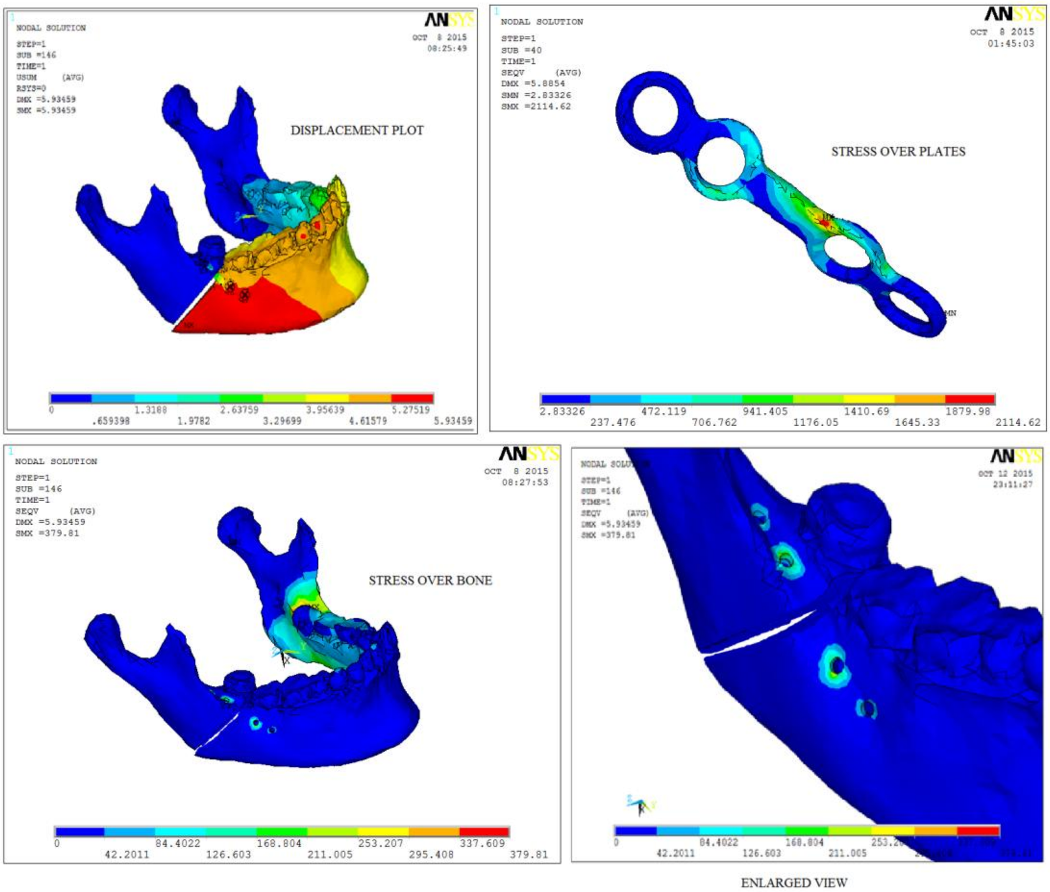


Figure 44: RED DENOTES MAXIMUM VALUE AND DARK BLUE DENOTES MINIMUM VALUE

RESULT INFERENCE:

Symphysis fracture fixation showed maximum deflection with Orthomax conventional and least with Leforte locking plates. Maximum stress over bone and plate was observed in Orthomax conventional and least with Synthes locking. Fracture gap was more in Orthomax conventional and least with Leforte locking. So the more stable plate is Synthes locking plate followed by Leforte locking plate.

Angle fracture fixation showed maximum deflection with Orthomax conventional and least with Synthes locking plates. Maximum stress over bone was with Orthomax conventional and least in Synthes locking plates. But stress over plate was more in Leforte conventional and least in Synthes locking. Fracture gap was more in Orthomax conventional and least with Leforte locking. So the more stable plate is Synthes locking plate followed by Leforte locking plate.

Hence, the dimensional changes in the holes of miniplates occurring during adaptation of the plate to the fracture site are also a factor for stability of the plate.

DISCUSSION

Normal mandible is subjected to bending forces in the upper border & to compression forces in the lower border.[1,2,3,4] In angle and symphysis fracture, torsional forces act in the angle region and tension forces are more concentrated in the symphyseal region.[10,11,12] Symphysis & angle fractures are 2nd most common fractures after condylar fractures. [11,17,28,29] The three main principles of fracture management include anatomical reduction, fixation and immobilization. The displacement of fracture is high in the symphysis and angle region, hence decrease in the bite force level. [12,30,31] This is because, in mandibular fracture there is a discontinuity and change of the stress trajectory.[5,32] Miniplates are designed in such a way to regain the stress trajectory continuity and counteract the forces.[5,33]

Numerous in-vitro and Finite Element studies on biomechanical evaluation of condylar fractures are available, but only limited in-vitro and Finite Element studies are available on symphysis and angle fracture of mandible. The elastic limit of flexibility of entire mandible is between 700 to 800 Mpa and rupture point is 950 to 1100 Mpa in mandibular fractures.[34]

In-vitro studies require expensive equipments and hence computerized studies evolved. FEA is one such study with great accuracy. FEA is a computational technique and most reliable analysis, originally developed by engineers, to model the mechanical behaviour of complex structures.[35,36] Biomechanical evaluation of mandible with simulated fractures, fixed with miniplates, are solved with ease.[21,25,37,38]

Dirk Vollmer etal did both experimental and FEA on mandible and concluded FEA is a powerful contemporary research tool.[7,28,39] Hence FEA is considered in our study to analyse stress pattern and biomechanical behaviour of 2.0mm Titanium four hole with gap miniplates namely Orthomax Conventional Plates, Leforte conventional, locking plates and Synthes locking plates in two different trauma situations namely angle and symphysis fractures.

Hideki Takada etal in his FEA study used dentate cadaveric human mandible with presence of third molar and suggested that presence of impacted or erupted third molars increased the risk of angle fracture by altering the concentration and transmission of stress in mandible.[23] Hence we also selected dentate human cadaveric mandible with completely erupted third molars for our study.

Atson Carlos de Souza Fernandes etal studied that the direct measurement and CT measurement of the cortical & cancellous bone thickness at the plating sites were almost the same. He proposed vestibular cortical bone should be between 3.0mm to 3.3mm for maximum stability of screws. Interradicular distance was greater than 2.0mm and hence the roots of the teeth were spared when screwed.[24] Hence, in our study, through CT scan the selected mandible was made sure that it had a cortical thickness similar to this study. This CT data in DICOM format was imported into MIMICS software and the computerized mandibular model was generated.

Stefano Benazzi etal studied the different elements of the mandible namely enamel, dentin, pulp, periodontal ligament using FEA. Each one was a separate element and had its own material properties, Young’s Modulus and Poisson’s ratio.[36]

Torreira M. G. and Fernandez J. R. analysed the material properties for compact and cancellous bone.[28] Baohui Ji etal analysed the material properties for Titanium plates and screws.[34] Hence our mandible’s and Titanium miniplates material properties were made sure that they had similar material properties like the above mentioned FEA studies.

Baohui Ji etal analysed that, when a fracture happens, the contacts between the fractured segments will be rough and serrated. To virtually simulated fracture site, they incorporated the contact as rough into the FE Study.[31,40] Hence, in our study, rough contacts were taken for the boundary conditions and fracture simulated in the angle and symphysis region.

Trivellato P. F. B. etal simulated the angle fracture with and without continuity of the inferior border of mandible and analysed the mechanical resistance of 2.0mm miniplates fixed with monocortical screws. Unfavourable angle fracture with or without discontinuity of the inferior border of the mandible did not decrease the mechanical resistance of the fixation of 2.0mm miniplates. Hence in our study we simulated unfavourable angle fracture of mandible.[41]

Julie Kimsal etal analysed Titanium as strong metallic element and it was more biocompatible with bone, which had least or no strain. Because of its high yield strength, dimensional changes of titanium miniplates were minimal, when bent and adapted. These dimensional changes could not be completely eliminated/avoided depending upon the site of fixation. Titanium was well understood as a good implant device as there was no need for plate removal at later stage.[42] Hence Titanium was the material of choice in our study.

According to Edward Ellis III 2.0mm Titanium miniplates were easy to adapt for any intraoral approach of any mandible fracture. But among the mandibular fractures, angle & symphysis plate fixation was more challenging next to condyle fractures. The more manipulation for adaptation of miniplates along the Champy's line of osteosynthesis happened in the angle region followed by symphysis region.[14] In our study, though manufacturer prescribed instruments were used for adaptation, the holes of titanium miniplates showed dimensional changes.

Righi E. etal developed plate Geometrics for the titanium miniplates which he used.[22] Similarly, we also developed the profile of the titanium miniplates using RAPID I Vision Measuring System (VMS). RAPID I-VMS could identify the diameter, radius, length and breadth of the miniplates by magnifying it.[43] Each miniplate, both conventional and locking plates were measured with high accuracy. The error level in this system was +/- 0.003mm. Each plate hole was measured pre and post bending and adapting to the angle and symphysis region of the mandible. The measurements like diameter and radius of the holes and change in centre were compared to the normal ones and the difference in between them were noted. This suggested that there are dimensional changes in the holes while adapting to symphysis and angle region. Plate-screw interface gap was present preventing perfect locking between plates and screws both in conventional and locking systems of the considered miniplates. Locking system of the miniplates had been designed in such a way by which the screws on placement are further held by second thread below the screw head-end in a way that the desired maximum stability was achieved. The conventional plates and screws interface each other with friction lock. Due to dimensional changes this interface was reduced resulting in decreased forces transmitted to the underlying bone. 3D scanning of the bent and adapted plates and screws were done through ATMOS scanner. CAD modelling software was used for model assembly. Hypermesh was used for fine meshing of the model and was solved in FEA.

Ribeiro-junior P.D. etal observed in their studies that the degree of plate adaptation affected the mechanical behaviour of non-locking plates but did not affect the locking plate-screw system. They also observed that in locking system the drill hole should be in the centre to facilitate proper screw locking to the plate.[44,45]

Yang L. and Patil P.M. etal demonstrated that 2.0mm locking miniplate/ screw system resulted in lower occurrence of complications and provided better stability of fracture fixation. Single 2.0mm locking miniplate placed along the superior border of the mandibular angle provided highly effective fixation for angle fractures without need for postoperative IMF.[46,47,48] Cheng-Jen Chuong etal analysed that 2.0mm miniplate with monocortical srews system has high mechanical stress when compared to other rigid fixation techniques but better when compare to non-rigid fixation techniques.[49,50,51,52] Hence in our study we compared 2.0mm titanium four hole with gap conventional and locking plates/screws systems in angle and symphysis fractures.

James W. Sikes etal compared the resistance to displacement between locking head and conventional screw types. Locking head screws provided superior resistance to conventional screws when two screws per segment was placed. No significant difference between the two types were found when all four screws were placed.[53] Hence, in our study, all four holes were fixed with their respective screws in the CAD model of fracture simulated mandible.

Osborn J.W., Baragar F.A., James W. DeVocht etal studied the pattern of human muscle activity and position of TMJ during clenching in 3D computerized model of jaw during vertical bite forces. 13 muscle elements for each side and a total of 26 muscle elements were evaluated. He categorised jaw muscles into Power muscles (Masseter, Medial Pterygoid and Temporalis) and control muscles (oblique fibres of Temporalis and Lateral Pterygoid). Power muscles cause displacement of condyle up and down and control muscles counteract this displacement. Hence the mandible is maintained in equilibrium and the condyle remained Static during Clenching (i.e.) during symmetric Vertical bite forces.[54,55,56,57] Kay-Uwe Feller etal established direction of vectors & forces for masticatory muscles. In his study the Lateral Pterygoid muscle consisted of two differently directed vectors.[26] Hence in our study we used muscle forces, direction vectors and muscle elements from the above mentioned studies for FEA.

Charles H. Gibbs etal evaluated the bite forces for normal individuals and found that it was as high as 443kg. The biting strength of natural teeth ranged from 55 to 280 lbs (25 to 127 kg) average of 162 lbs (74 kg) in his study.[58] Proffit W.R., Fields H.W., Nixon W.L. evaluated occlusal forces during swallow, simulated chewing, and maximum effort in 19 long-face and 21 normal individuals using both quartz and foil based piezoelectric force transducers. Forces were measured at 2.5 mm and 6.0 mm molar separation. Long-face individuals have significantly less occlusal force during maximum effort, simulated chewing, and swallowing than do individuals with normal vertical facial dimensions. No differences in forces between 2.5 and 6.0 mm jaw separation were observed for either group.[59] Gerlach K.L. etal evaluated maximal unilateral biting forces (fracture side) in 22 patients with mandibular angle fractures treated with miniplate fixation according to Champy's line of osteosynthesis. This revealed that after fixation 1 week postoperatively only 31% of the maximal vertical loading found in controls was registered. These values increased to 58% at the 6th week postoperatively.[13]

In our study maximal vertical bite forces for control groups with 10 patients were evaluated for 5 bite points – the right molars, right premolars, anteriors, left premolars and left molars. The vertical bite forces were evaluated using an electronic bite force gauge. Pain being the only limiting factor the bite forces were evaluated pre op, 1 week, 1 month and 2 months post operatively for the fracture group. 2 months postoperative maximal vertical bite forces for 8 patients treated with symphysis fracture and 7 patients treated with angle fracture with titanium miniplates according to Champy's line of osteosynthesis were evaluated. Maximal vertical bite forces in the right molar and premolar region was 42.41kg (avg), left molar and premolar region was 48.07kg (avg) and anterior region was 13.72kg (avg). These values were incorporated in the FEA for the amount of force to be applied in the model.

Hasan Husnu Korkmaz evaluated the deflection and stability of miniplates and Von Mises stress over bone and miniplates. In addition to this, he also measured the fracture gap for different plating systems in his FEA on mandibular fractures.[21]

Ziebowicz A. etal studied the upper limit of dislocations of the blocks of a broken mandible in the gap of fracture site. Under the applied forces the fracture gap did not exceed 1 mm in the Champy's technique.[32,60] Tsutomu Sugiura etal reported that proper healing required a maximum displacement of less than 150 µm at the fracture site. [29,44,61] Analysis of fracture gap in our study showed that maximum fracture gap was identified in the Orthomax conventional miniplates for angle fracture being 2.2708 mm (Table – 13) and for symphysis fracture being 0.86241mm (Table – 13). Locking miniplate osteosynthesis systems used in symphysis and angle fracture were ranging between 0.09mm to 1.29mm (Table – 13) respectively. This shows that the Orthomax plates were great fixation failure compared to the other ones.

Symphysis fracture fixation showed maximum deflection of 6.05196mm (Table – 10) with Orthomax conventional and least of 2.50747mm (Table – 10) with Leforte locking miniplates. Maximum stress over bone was 98.6587 MPa (Table – 11) with orthomax conventional and least was with synthes locking of about 78.476 MPa (Table – 11). Stress over plate was more of about 75.4011 MPa (Table – 12) in Orthomax conventional and least of about 61.2447 MPa (Table – 12) in Synthes locking. Fracture gap was more of about 0.86241mm (Table – 13) in Orthomax conventional and least of about 0.01804mm (Table – 13) with Leforte locking. So the more stable plate is Synthes locking plate followed by Leforte locking plate.

Angle fracture fixation showed maximum deflection of 5.93459mm (Table – 10) with Orthomax conventional and least of about 3.00287mm (Table – 10) with Synthes locking plates. Maximum stress over bone was more of about 379.81 MPa (Table – 11) for Orthomax conventional and least of about 309.63 MPa (Table – 11) for Synthes locking plates. Stress over plate was more of about 2114.62 MPa (Table – 12) in Orthomax conventional and least of about 833.457 MPa (Table – 12) in Synthes locking. Fracture gap was more of about 2.2708mm (Table – 13) in Orthomax conventional and least of about 1.86241mm (Table – 13) with Leforte locking.

So the more stable plate is Synthes locking plate followed by Leforte locking plate. The locking plates had better stability than any other. Hence the dimensional changes in the holes of miniplates occurring, during adaptation of the plate to the fracture site is also a factor for the stability of the plate.

SUMMARY & CONCLUSION

The modern traumatology demands that we evaluate the risk and benefits of each treatment modality and apply it appropriately for each patient. As treatment options expand, it remains imperative that we consider the anatomic, physiologic and biomechanical factors associated with managing these injuries.

The advent of osteosynthesis in maxillofacial surgery developed different system and designs to overcome the shortcomings in the existing devices. In the present study conducted at the department of oral and maxillofacial surgery, Sri Ramakrishna Dental College and Hospital, an attempt was made to identify the factors influencing the miniplate fixation in angle and symphysis fracture of mandible.

From our study it is found that 2.0mm titanium 4 hole with gap conventional and locking plate/screw system irrespective of the manufacturer shows dimensional changes in the holes when adapted to angle and symphysis region. Because of these dimensional changes the respective plate screw interface is not intact leading to altered stability, increased fracture gap and increased stress over plates and uneven stress over the bone. Hence dimensional changes of the holes are also one among the factors to be considered during plate fixation in angle and symphysis region.

The study on stress distribution and distortion proved to be the cause for minor alteration in the outcome of the stable occlusion. Anyhow the results have to be applied clinically as it is in-vitro study. The clinical situations are much more complex than the mechanical simulation and the evaluation on different situation will provide much more knowledge on the SRIF plating system.

DECLARATION OF CONFLICT OF INTEREST:

I hereby declare that no person or organization is involved in multiple interests or financial interests in the entire study.

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ANNEXURES

ANNEXURE I – PATIENT DATA

S.No	PATIENT NAME	AGE	SEX	CAUSE OF INJURY	DIAGNOSIS	FIXATION
1	Mr. Jagadeesh	26	M	RTA	# Symphysis	Titanium
2	Mr. Velusamy	50	M	RTA	# Right Angle	Titanium
3	Mr. Gowthaman	22	M	RTA	# Symphysis	Titanium
4	Mr. Kanagaraj	30	M	RTA	# Right Angle	Titanium
5	Mr. Kamesh	21	M	RTA	# Right Angle	Titanium
6	Mr. Sundar	30	M	RTA	# Symphysis	Titanium
7	Mr. Sampath Kumar	26	M	RTA	# Symphysis	Titanium
8	Mr. Palanisamy	26	M	RTA	# Right Angle	Titanium
9	Mr. Ravi	33	M	RTA	# Symphysis	Titanium
10	Mr. Raghuveer	24	M	RTA	# Symphysis	Titanium
11	Mr. Santosh	28	M	RTA	# Symphysis	Titanium
12	Mr. Satish Kumar	24	M	RTA	# Symphysis	Titanium
13	Mr. David	19	M	RTA	# Right Angle	Titanium
14	Mr. Sakthivel	25	M	RTA	# Symphysis	Titanium
15	Mr. Vijayaraghavan	34	M	RTA	# Symphysis	Titanium

ANNEXURE II: BITE FORCES IN CONTROL GROUPS

S.NO	AGE	MOLARS (RIGHT) (Kg)	PREMOLAR (RIGHT) (Kg)	ANTERIOR TEETH (Kg)	PREMOLAR (LEFT) (Kg)	MOLARS (LEFT) (Kg)
1	20	31.699	12.491	16.329	6.695	24.18
2	22	20.246	4.397	12.491	5.526	21.326
3	25	34.177	15.998	12.931	13.381	22.345
4	30	18.677	7.004	11.402	6.485	19.097
5	32	18.876	1.988	12.012	2.588	19.167
6	38	28.280	3.897	9.393	3.398	18.487
7	43	34.336	10.713	20.645	7.025	23.344
8	45	32.378	11.192	14.390	10.693	39.173
9	47	28.089	10.392	8.314	10.392	21.814
10	50	17.567	4.407	10.193	4.477	22.874

ANNEXURE III: BITE FORCES IN OPERATED GROUPS – 2 MONTHS POST OP

S.No	PATIENT NAME	AGE	SEX	MOLARS (RIGHT) (Kg)	PREMOLAR (RIGHT) (Kg)	ANTERIOR TEETH (Kg)	PREMOLAR (LEFT) (Kg)	MOLARS (LEFT) (Kg)
1	Mr. Jagadeesh	26	M	14.162	12.183	4.863	16.124	13.189
2	Mr. Velusamy	50	M	16.612	14.004	4.628	15.592	17.013
3	Mr. Gowthaman	22	M	16.632	17.062	10.162	15.898	19.021
4	Mr. Kanagaraj	30	M	20.173	20.067	12.632	24.689	30.156
5	Mr. Kamesh	21	M	25.637	20.563	9.145	30.642	32.002
6	Mr. Sundar	30	M	15.637	14.682	11.113	20.511	22.600
7	Mr. Sampath Kumar	26	M	25.129	26.002	20.420	35.620	40.002
8	Mr. Palanisamy	26	M	17.182	13.666	12.524	15.125	17.963
9	Mr. Ravi	33	M	26.289	15.412	15.043	18.616	25.824
10	Mr. Raghuveer	24	M	32.632	25.600	13.615	26.501	31.987
11	Mr. Santosh	28	M	21.145	20.001	15.842	19.637	25.624
12	Mr. Satish Kumar	24	M	17.733	14.615	12.455	15.540	20.155
13	Mr. David	19	M	25.542	23.655	20.166	26.744	27.741
14	Mr. Sakthivel	25	M	35.630	31.154	21.615	27.754	29.984
15	Mr. Vijayaraghavan	34	M	29.656	27.753	21.636	27.762	31.004

ANNEXURE - IV

CASE HISTORY PROFORMA

Name: Age: Sex:

Hospital no:

Address: Occupation:

Contact no:

Ip no:

DOA:

DOD:

DOS:

History of trauma:

History of Present illness:

History of LOC:

Bleeding from ear:

Bleeding from nose:

Bleeding from oral cavity:

Medical History:

Personal history:

Family history:

Past Dental History:

Drug Allergy:

General physical Examination:

Built:

Nourishment:

Gait:

Vital signs:

BP:

Pulse:

Temperature:

Respiratory rate:

Extra oral examination:

Inspection:

Face: symmetrical / Asymmetrical

Left side:

Right side:

Subconjuntival haemorrhage

Palpation:

Pain/tenderness:

TMJ:

Clicking:

Deviation:

Tenderness:

Movements:

1. Lateral

2. Protrusive

3. Retrusive

Lymph node:

Intra oral examination:

No of teeth present:

Tenderness / mobility of fracture segments:

Step deformity of the mandible:

Occlusal derangement:

Bite force registration-right molar:

Left molar:

Incisor:

Provisional Diagnosis:

Investigations: CT, OPG, blood

OPG findings:

Blood investigations:

Blood group:

Urea:

Creatinine:

RBS:

ESR 1 hour:

WBC TC-

N- , L- , E-

Platelet- lack cells /cu mm

RBC count- million cells /cu mm

Hb: gms%

BT: min, CT- min

Pre anaesthetic fitness:

Final diagnosis:

Treatment plan:

Treatment done:

Review: Occlusion

Right molar Left molar

Bite force measurements 7th day

3rd week

6th week

3rd month

Post-operative evaluation of the patient:

Inclusion criteria:

- Post-operative pain
- Mouth opening
- Occlusion
- Bite force measurement

Exclusion criteria:

- Paresthesia
- Post-operative infection
- Oedema
- Step deformity
- Malunion
- Patient with systemic illness
- Patient who are not willing for surgery
- Patient with no proper follow up