Three dimensional modelling and finite element distortion analysis of the mandible

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THREE DIMENSIONAL MODELLING AND FINITE ELEMENT DISTORTION ANALYSIS OF THE MANDIBLE

by

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A dissertation submitted in fulfillment of the requirements
for the degree of Doctor of Philosophy at
The University of New South Wales.

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"I hereby declare that this submission is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person nor material which to a substantial extent has been accepted for the award of any other degree or diploma of a university or other institute of higher learning, except where due acknowledgement is made in the text."

ABSTRACT

In the past due to the complexity of its movements, the biomechanics of the mandible have proved difficult to analyze. Even tough estimation of the temporomandibular forces from mathematical models has a long history , it has led to quite conflicting results.

Until to-day, only few papers have been published concerning the development, analysis and experimental verification of three dimensional, finite element models of the human mandible. The mathematical models used in these studies have been almost exclusively first-order, partially completed, over constrained structural models, exhibiting limited anatomic description and properties. Since structures in oral cavity are complex by nature the only useful mathematical models of the mandibular environment are the ones which represent these complexities. Improvement on previous models should include, forces acting on the mandible, realistic geometry and boundary conditions analogous to musculature support and ultimately better bone and tissue property characterization both structurally and biologically. The use of computers are essential if the analysis is to be applied to physiologic problems.

In this thesis a three-dimensional finite element model of the mandible in its intact, physiologic state has been developed using CAD methods. Both cortical and cancellous bones were represented in the 3-D model. The model geometry was derived from physical measurements taken of an average size mandible. Data on anatomic and functional aspects, and material properties were obtained from literature. Materials were idealized as homogeneous, isotropic and linearly elastic. The complete model consisted of 258 (20 node) solid elements and 1106 nodes.

Three nodes on the symmetry plane were restrained. Reactions of those points were shown to be zero so that constraining these parts had no effect on the stress distribution. Thus the model was completely free to deform, removing the approximation caused by over-restraining in previous models.

It has been reported that the mandibular bone flexes when subjected to a variety of intrinsic and extrinsic forces. A survey of the literature reveals that the possibility of the mandibular distortion during functional movements has not been biomechanically investigated and to date no mathematical model of the mandible has been attempted to solve this or other related problem.

The distortion of the mandible by using two different three dimensional models has been found to occur.

When published values of distortions are compared with this work, the calculated values are in good agreement with the previously measured values.

The range of mandibular movement and its distortion during clenching, opening and protrusion is dependent on mandibular geometry, mechanical properties and the thickness of the cortical bone, and the type, magnitude and to the point of application of the muscle forces.

The range of mandibular distortion during opening and protrusion could present some important clinical problems.

Several improvements on the development of the 3-D model also have been suggested to provide better insight in the fields of biomechanics, oral implantology, periodontology, prosthetic dentistry, maxillofacial reconstructive surgery, temporomandibular joint dysfunction and periodontic corrective device methodology.

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LIST OF SYMBOLS

a_b	Moment arm of the bite force
a_m	Moment arm of an muscle
a_n	Moment arm of the n th muscle
b_k	Moment arm of the k th bite force
[B]	Partial derivative of the matrix [N]
B_k	The k th bite force
B	Bite Force
$[B]^T$	The transpose of matrix [B]
c_q	Moment arm of the q th joint force
[D]	Elasticity matrix for the material of the element
E_1	Young's modulus value of the cortical bone
E_2	Young's modulus value of the cancellous bone
E_{can}	Young's modulus value of the cancellous bone
E_{cor}	Young's modulus value of the cortical bone
F_b	Bite force
F_E	Muscle Force (Lateral Pterygoid)
F_I	Muscle Force (Medial Pterygoid)
F_j	Joint force
F_M	Muscle Force (Masseter)
F_m	Muscle force
$F_m(max)$	Physiological maximum force
F_{mp}	Medial pterygoid muscle force
F_n	The force of the n th muscle
F_p	Lateral pterygoid muscle force
F_T	Muscle Force (Temporalis)
F_x	Force component in x axis
F_y	Force component in y axis
F_z	Force component in z axis
J_q	The q th joint force
[K]	Overall stiffness matrix
K	Number of bite forces
$[k]^e$	The stiffness matrix for an element
L	Magnitude of load
Lp	Lateral Pterygoid muscle
M	Masseter muscle
M_1	First molar
M_2	Second molar
M_3	Third molar

Mp	Medial Pterygoid muscle
N	Number of muscles
[N]	A matrix of shape function
O	Opener (digastric, mylohyoid, geniohyoid) muscles
[P]	Overall load vector
$[P]^e$	Load vector for an element
P_1	First premolar
P_{My}	y component of masseter muscle
P_{Mpy}	y component of medial pterygoid muscle
P_x	x component of the muscle force
P_{xy}	Projection of the force P on a xy co-ordinate plane
P_{xz}	Projection of the force P on a xz co-ordinate plane
P_y	y component of the muscle force
P_z	z component of the muscle force
P_{zy}	Projection of the force P on a zy co-ordinate plane
Q	Number of joint forces
R	Temporomandibular joint force resultant
T	Temporalis muscle
ta	Anterior temporalis muscle
tp	Posterior temporalis muscle
[U]	Displacement
$[u]^e$	The displacement of the nodes of an element
α_k	The angle of the k th force to y axis
α_m	The angle of the muscle force to y axis
α_n	The angle of the n th force to y axis
α_q	The angle of the q th force to y axis
β_k	The angle of the k th force to x axis
β_n	The angle of the n th force to x axis
β_q	The angle of the q th force to x axis
Υ_k	The angle of the k th force to z axis
Υ_n	The angle of the n th force to z axis
Υ_q	The angle of the q th force to z axis
Φ_m	Physiological cross section
Γ	Maximum muscle tension
Φ	Joint force resultant angle to z axis
θ	Load angle to z axis
$[\sigma]$	Stress matrix
$[\epsilon]$	Strain matrix

INTRODUCTION

1.1 Mandibular flexure and distortion.

The mandibular bone flexes when subjected to a variety of intrinsic and extrinsic forces. This response is a function of the mechanical properties of the bone as well as the type, magnitude, direction and point of application of the force.

For many years from the clinical standpoint the bone was thought to be rigid and unyielding. The first suggestion that the medial angulation of the external pterygoid muscles could introduce a contracting action upon the two halves of the jaw was made by Grunewald (1921). The effect of the mechanical deformation on the mandible was studied by Kuntscher (1934) in vitro, using colophonium and by Evans (1953) and Du Brull and Sicher (1954) with a "stress-coat" technique. Evans studied the strain due to statically applied loads of the order of 260 N applied to the chin and to the ramus near the angle. Du Brul and Sicher examined the effect of manual force tending to squeeze the condyles together. They reported stress cracks near the symphysis and at the point of insertion of the lateral pterygoid muscles. They demonstrated that the lateral pterygoid muscles contracting in an almost frontal plane during opening and protrusion of the mandible pull the condyles together. This contraction causes flexure, around the mandibular symphysis, with a resultant sagittal movement of the posterior segments. They have reported a distortion between the last mandibular molars which was up to 0.5 mm less when the jaw was forcefully protruded than when it was at rest.

Distortion of the mandible in vivo due to physiological forces was detected by Jung (1952, 1959). The method employed a modified dial gauge which measured the distance between the lingual surfaces of contralateral teeth. When the subject's mouth was opened, the mandible was moved laterally, or closure was attempted against resistance, a reduction in the inter-premolar and inter-molar distance was detected. The lower jaw was found to be distorted more than the upper jaw. In the mandible a greater reduction in width was recorded in the molar than in the premolar region. These findings were considered to be due to distortion of the jaws caused by contraction of the muscles of mastication and of the floor of the mouth.

Weinmann and Sicher (1955) also have stated that "The bending force is exerted mainly by the medial component of force of the obliquely arranged external pterygoid muscles. If the two external pterygoids are forcefully contracted the mandibular condyles are pulled medially and the mandible is measurably deformed".

Most earlier studies of mandibular arch deformation in vivo were performed by using occlusal templates and micrometers.

McDowell and Regli (1961) determined arch width changes by securing intra oral metal splints using measuring points in the first premolar and second molar regions. Only the "maximal readings" were recorded and averaged. "Minimal dimensional changes" were observed in the premolar region. In the 20 patients studied, the average dimensional change was reported to be from rest to wide open 0.4 mm and in protrusion 0.5 mm. Also maximum recorded change of 1.5 mm decrease in width was reported in the forced protrusive position.

Most studies of the effect of force to the skull as a whole have been concerned with the results of injuries so that only static loads have been used and the amount of force has been large.

Dynamic distortion of the bone due to varying loads was studied in the skull of dogs by Gurdjian and Lissner (1944) who used wire resistance strain gauges. The same principle was used by Evans (1953) to detect physiological distortion of the tibia of a dog during walking. Force applied to the teeth was found, by Muhlemann (1954), to cause elastic deformation of the alveolar margin in Rhesus monkeys. Static buccolingual force was used. Force above 100 gr produced displacement of the bone detected by a dial gauge anchored to the teeth.

Movement of the tooth caused tension of the principal fibres of the periodontal membrane on the side of the applied force. Progressive deformation of the bone in this region was observed as the force increased.

Zwirner (1949, 1951) reported a method utilizing a movable plates of a condenser, one attached to the tooth to be measured and the other to the substance in which the jaws of a rat were embedded.

Picton (1962) described a clinical method of measuring axial tooth movement by means of resistance-strain gauges. Biting force was exerted on the teeth through a dynamometer employing a further wire strain gauge. Fourteen young male subjects were studied. For the test, the rubber dam clamp carrying two transducers were mounted on the first or second premolar while the dynamometer was placed between the teeth on the opposite side of the mouth so that force could be applied to the contralateral tooth. He has reported that when biting force is applied to a tooth, the tooth is depressed into the socket and tilts mesially producing slight mesial displacement of the anterior teeth. At the same time distortion of the jaws occurs which may spread to cause movement between adjacent teeth on the contralateral side. He further concluded that this contralateral deformation is due to distortion of the mandible.

Osborn and Tomlin (1964) developed a different method to measure mandibular distortion using an intraoral transducer suspended between mandibular molars. Measurements were carried out on 40 people, the average medial movements reported, on opening was 0.07 ± 0.013 mm and on protrusion 0.09 ± 0.013 mm. They haven't found any statistical evidence showing that the amount of medial movement resulting from opening is greater than resulting from protrusion. Neither was there any evidence of the amount of medial movement (on opening or protrusion) is related to the maximum extent to which the mouth could be opened. There appeared to be no correlation between the age of the subject and the degree of flexion. There is little doubt that a properly calibrated variable capacitance transducer could be made to yield accurate results, but the bulk of the one used in this project appeared to occupy tongue space. In keeping the tongue from containing the transducer, the subject may have experienced some limitation of mandibular movement.

Regli and Kelly (1967) found a decrease in mandibular width during opening at the second molar region. Using occlusal templates of cold curing acrylic resin and a micrometer, measurements were made on casts taken at various vertical openings. They have reported average distortion across the mandibular arch from first bicuspid to first bicuspid to be 0.03 mm and that from second molar to second molar 0.09 mm. The greatest change in dimension, 0.2 mm was measured at the second molar level in one subject.

Burch and Borchers (1970) designed a device to record the process of change in mandibular arch width as the mandible moved through its various excursions. They used strain gauges, mounted above and below a beryllium-copper spring steel strip, slightly bowed vertically across the mandibular arch and adapted to fit between the right and left first molars. Ten subjects were tested, they showed decrease in mandibular arch width during movement from rest position to a full protracted, a wide open, a right lateral, and a left lateral position. They have reported the mean magnitude of decrease in arch width of 0.610 mm in protraction, 0.438 mm in opening, 0.243 mm in right lateral, and 0.257 mm in left lateral movements.

Bowman (1970) also used strain gauges to evaluate flexure of the mandible between the first molars. The mean medial flexure was 0.29 mm for maximum opening and 0.637 mm for full protrusion. Bowman also observed that adduction continued following an opening movement of 20 mm. Adduction during protrusion began immediately and was proportional to the amount of protrusion. The effects of incisal biting and horizontal retruding force caused the mandible to flex laterally an average of 0.27 and 0.18 mm respectively.

Novak (1972) conducted three separate experiments, the first part was carried out using an intraoral device capable of measuring the distance between the mandibular second molars, second part consisted of making a wax intraoral occlusal record and cast of the lower arch. The third consisted of having intraoral occlusal films taken of a mandible in an open and closed positions. He reported that mandible flexed inward 1 mm in the first molar region when the mandible was in a wide open position. He further concluded that in the region of the angle, this flexion could be greater.

In 1972, Burch designed a device that could record the dynamics of mandibular arch width change. A strip of beryllium copper, 6.5 mm wide and 0.21 mm thick was shaped to form a vertical arch, and two strain gauges were cemented to it above and below the greatest curvature of the arch. The instrument was placed between the jaws of a micrometer caliper and coupled to a polygraph for calibration and testing. The device was then used on 25 dentists and students to record changes in arch width. The apparatus was affixed to mandibular first molars and coupled to a polygraph while the subject was instructed to perform various mandibular movements. Wide opening and protraction from rest position produced 0.224 mm and 0.432 mm decrease in arch width respectively. It was further reported that maximum decrease was not maintained when the fully protracted or open positions were held for 6 to 10 seconds.

Goodkind and Heringlake (1973) attached a metal clutch with a micrometer to the posterior teeth of 20 male and female students. The projecting arm of the clutch afforded extra oral anchorage of the dial micrometer. The gauge's probe was directed against the buccal surface at the height of contour of the second molar tooth in the contralateral mandibular quadrant. The mean premolar region values reported was 0.0316 mm and for the second molar region was 0.0768 mm. They have also noticed a continuous change in the amount of flexure with a brief stabilization at maximum opening. Upon closing the arch width dimension was found to return to its original value as recorded by a zero gauge reading.

De Marco and Paine (1974) designed a gauge which could be placed on the occlusal surfaces of the mandibular first molars. Seven men and 18 women were subjects in this study. The amount of maximum opening was determined by measuring from the base of the nose to a mark placed on the chin. This was measured with the teeth in centric occlusion and at maximum opening. The difference between these two measurements was called the maximum interocclusal distance. All measurements were expressed as percentages. In the 25 patients tested, the average maxillo mandibular opening was found to be 52.5 mm, while the mandible was found to contract an average of 0.78 $\pm$ .05 mm during maximum opening. A range of 0.6 to 1.5 mm was reported. The results were plotted and it was noted that there was no change in the width of the mandible up to 28 percent opening. Thereafter, the dimensional change that occurred was directly related linearly to the percent of opening.

The effect of fixed prostheses or mandibular arch width in a forced open position was investigated by Fischman (1976). With various combinations of fixed splints in place, changes in the width of the mandibular arch during forced opening were measured. Cast copings with receptacles for the measuring caliper were constructed for the mandibular right first and left second molars. Measurements were made in the most closed and forced open positions. The amount of control flexure was reported to be 0.86 mm (0.0339"). In all instances, mandibular flexure was reported to be reduced when fixed splints were present in the mouth. He also reported that none of the splints tested completely inhibited flexure. From this he concluded that all splints undergo at least some form of stress and in some cases flexure during mandibular opening.

Omar and Wise (1981) measured the mandibular flexure in the horizontal plane, with an "anterior jig" when the retruded axis position (R.A.P.) recordings were made, during chin-point guidance and patient applied muscle forces. The retruded axis position is considered by many authors to be the relationship of the mandible to the maxilla when the condyles are most posteriorly and superiorly located in the glenoid fossal (Boucher, 1974). Many workers agree that the reproductibility of this position renders it a valuable reference position (Kantor, Silvermann and Garfinkel, 1972). Other research workers have shown that the R.A.P. exhibits variability. Variations in the R.A.P. have shown to be time-dependent (Shapagh, Yoder and Thayer, 1975), posture dependent (Helkimo, Ingervall and Carlsson, 1973) and technique dependent (Kantor et al., 1972; Strohaver, 1972; Calagna, Silvermann and Garfinkel, 1973; Federick, Pameijer and Stallard, 1974). In Omar and Wise's studies ten subjects were investigated, mandibular horizontal flexure was observed in all subjects during a patient-applied muscle force R.A.P. registration. In all cases there was an increase in mandibular arch-width. The mean increase was 0.073 $\pm$ 0.028 mm. A decrease in mandibular arch width in wide opening movements has occurred in all cases. A mean of 0.093$\pm$ 0.044 mm, with a range of 0.012 mm to 0.164 mm, was reported.

The range of movement, was a mean of 0.166 mm, although on an individual basis, the range was 0.084 mm to 0.234 mm.

The changes of the mandibular arch width and distortion by using light sensing photodiode (L.S.P.) was carried out by Gates and Nicholls (1981). Positional changes of the light on the diode surface were detected and quantitatively measured by relating output voltage to light source. This photo diode-fibreoptic system evaluated mandibular arch width changes in vivo. To compare arch width changes with mandibular opening and protrusion, a linear variable differential transformer (L.V.D.T.) mounted extraorally measured the quantity of the opening or protrusion while the photo diode has measured the mandibular arch width changes. They reported mandibular distortion values for opening which ranged from 0.053 mm to 0.306 mm and protrusion values ranged from 0.113 to 0.51 mm. They have concluded that when distortion was compared to millimeter of opening, the arch width change took place after 20 mm of opening which was in good agreement with Bowman's (1970) results. However, a linear relationship was not observed, after the patient exceeded 28 percent of maximal opening as reported by DeMarco and Paine (1974).

1.2 The Purpose of the Research:

The importance of understanding the human musculoskeletal system, which combines bones, joints and muscles in an intricate supportive and protective structure is of paramount importance in the field of reconstructive surgery and prosthetic replacement where the principal aim is to restore or replace a damaged tissue or organ so that the patient can function effectively.

The accompanying field of applied mechanics relies heavily on a proper understanding of anatomic systems before research methods can be applied and mathematical tools used, to gain insight into their physical nature.

The concern of this thesis is to examine the mandible, which as a biomechanical system, together with the temporomandibular joint, is one of the most complex of all anatomic systems. A fundamental insight into the biomechanics behaviour of this system and the functional aspects, is a necessity if appropriate surgical techniques and prosthetic devices are to be further developed.

In the past the main areas of study into the mandibular distortion have been centred on a simple type of investigation, in which the investigator has recorded the amount of distortion with various methods and equipment in bicuspid and molar areas of the mandible. To date no mathematical model of the mandible has been attempted to solve this or other related problems.

There were various reports on two dimensional and three dimensional modelling of the mandible. These models appear to be very remote from the actual anatomic and biomechanical conditions.

The purpose of this study was to improve the method of three-dimensional modelling and uniting biomechanics and medical science research approach by utilizing the model in an important maxillofacial problem namely "functional distortion of the mandible".

The degree of adduction of the mandible, when it is depressed or protruded, has been measured by a number of researchers (TABLE I).

TABLE I

The average and maximum degree of adduction (mm) of the mandible in first and second molars measured by various investigators during wide open and protrusion.

		OPEN				PROTRUSION			
		1ST MOLAR		2ND MOLAR		1ST MOLAR		2ND MOLAR	
	DATE	AVER.	MAX	AVER.	MAX.	AVER.	MAX.	AVER.	MAX
JUNG V. F.(1)	1952			0.35	0.84			0.70	1.00
DEBRUL & SICHER(2)	1954							0.5	
McDOWELL & REGLI(3)	1961			0.421	1.4			0.535	1.5
OSBORN & TOMLIN(4)	1964			0.07	0.33			0.09	0.46
REGLI & KELLY(5)	1967			0.09	0.18				
BURCH & BORCHERS(6)	1970	0.438				0.61			
BOWMAN A.(7)	1970			0.297	0.59			0.637	1.286
NOVAK C. A.(8)	1972	1.00							
BURCH J. G.(9)	1972	0.224	0.61			0.432	0.75		
GOODKIND& HERIN.(10)	1973	0.076	0.109						
DEMARCO & PAINE(11)	1974	0.78	1.5						
FISCHMAN B. M.(12) *	1976			0.861	1.11				
KOLLNER H. J.(13) **	1978	0.105	0.2						
OMAR & WISE(14)	1981	0.093	0.164						
GATES & NICHOLLS(15)	1981	0.142	0.306			0.29	0.51		

* Measured between mandibular right second and left first molars.

** Measured between the first and second molar interfaces.

There is considerable discrepancy between the results obtained by these researchers, but certainly some of the reported distortions are very significant. If these distortions really occur - as reported -, this must be taken into consideration by the dentists, surgeons and dental appliance designers for the following reasons :

i) The mandible may flex during impression making. A lower prosthesis fabricated from this impression may not seat.

ii) Modern rehabilitation concepts call for extreme accuracy in impression technique and in making inter-occlusal records. Flexion of the mandible during these critical recording procedures may create serious errors.

iii) Forced depression or protrusion of the mandible, in the presence of a removable or fixed partial denture, may cause undue stress on the abutment teeth or prosthesis.

iv) Patients with full lower dentures may also be affected by mandibular distortion, both by the initial fit of the denture resulting from the impression, and by the functional bending of the mandible while wearing the denture.

v) The effect of the mandibular distortion on the endosteal implant micromovements and osteolysis.

Experimental work and clinical experience in the field of rigid internal fixation in general and in mandibular fractures in particular provide a basis for the development of practical implant prosthetics.

It is apparent from clinical experience and from experiments that, the incorporation of a transplant and induction of bone formation succeed in locations where mechanical motion is absent. The theoretical principle, on which the insertion of an implant to secure the prosthesis is based, is that bone is not an "inert" supporting substance. Injury to bone always elicits the same chain of reactions, regardless of whether due to fracture or mechanical damage caused by the insertion of an implant: rapid proliferation of osteoblasts and osteoclasts in the endosteum and periosteum. The process is based on the principle of inducing osteogenesis. However, the formation of such tissue depends on the type of mechanical strain (Krompecher, 1937; Pauwells, 1965; Basset, 1962; Basset et al., 1961; Schenk, 1975). Bone forms directly only when mechanical movement is absent. By contrast, shearing forces stimulate the formation of cartilaginous tissue, and tractional forces the formation of all type of (bone, scar, cartiloginous) of connective tissue. Thus, mechanically induced reaction in the bone must be taken into account in the implantation of any prosthetic material.

In 1962 Basset postulated that, under certain conditions, one category of cells, the undifferentiated mesenchymal cells can apparently produce different types of tissue. The possibility of such cell modulation (translation from one functional form to another) is verified most strikingly in the use of endosteal implants.

During various mandibular movements, implants (the screw, needle or blade type) could be subjected to an alternating load oscillating through zero. The results are micromovements between the implant and the bone, which do not lead to osteogenetic but rather to fibroblastic differentiation. The proliferative connective tissue forms an additional barrier against the possible advance of angiogenic (vascular) bone tissue. The micromovements cause an increasing loosening of the implant. This condition and subsequent fibrosis, greatly facilitates infection. Resorptive granulation tissue is formed and destroys the contact zones between the implant and the bone. This process is designated as osteolysis by Ganz et al. (1975).

This osteolysis, which is induced by movement, and the greatly potentiated risk of infection could be the main cause of the failures recorded so far in dental implantology.

The dentists are also regularly confronted with patients who show symptoms of the mandibular dysfunction syndrome (Agerberg and Carlsson, 1972). According to Nordh (1974), the clinical characteristics of this syndrome are as follows:

i) Spasm of the elevators of the mandible;
ii) Pain in the head and face;
iii) Discomfort in the mandibular joints;
iv) Limitation in the movement of the mandible; and
v) Disturbances in occlusion and/or articulation.

Effective management of temporomandibular disorders depends on precise identification of the kind of dysfunction present in the complaint, where it is located, and what chiefly is responsible for it.

In patients with orofacial pain and functional disorders electromyography has been used to demonstrate abnormal muscle activity in the elevators of the mandible (Jarabak, 1957; Perry, 1955, 1957; Ramfjord and Ash, 1971; Moller, 1966, 1970). By quantitating the electrical activity they have shown that pain and tenderness of the same muscles are accompanied by increased postural activity and low maximal strength.

If the abnormal muscle activity is a source of pain in functional disorders of the masticatory system, treatment followed by the relief of pain must have involve changes in the action of the muscles of mastication.

Depending on their clinical similarities, temporomandibular disorders may be grouped into five classes proposed by Bell (1982):

i) Acute muscle disorders;
ii) Disc-interference disorders of the joint;
iii) Inflammatory disorders of the joint;
iv) Chronic mandibular hypomobilities
v) Growth disorders of the joint.

These functional disturbances of the masticatory system can be, as complicated as the system itself. Although numerous treatments have been advocated, none are effective for every patient every time. Effective treatment selection should begin with a thorough understanding of the disorder and its etiology. An appreciation of the various types of bones, muscles, ligaments and biomechanics of the temperomandibular joint is essential for understanding and effective managements of the symptoms. The anatomy and function of the temperomandibular joint have been well described in detail, in many textbooks. However until recently, very little information could be found about the changes in the joints that occur during normal development or associated with abnormal function or pathological conditions. A realistic three dimensional model of the mandible and temperomandibular joint might help to establish, specific treatment goals before the therapy begins.

Due to the complexity of movements, the biomechanics of the mandible have proved difficult to analyze. Even that estimation of the temperomandibular forces from two-dimensional mathematical models has a long history it has lead to quite conflicting results. The greatest controversy has been over whether the temperomandibular joint is even a load bearing joint or not.

With an exception of few 3-Dimensional partial models, most of the mandibular models are two-dimensional, two/four muscle models. The predictions made from such models are incomplete because the forces and movements operate in three dimensions and their projections onto a plane suppress what may be important biomechanical aspects of the system.

A survey of the literature suggests that the possibility of the mandibular distortion, during functional movements, has not been biomechanically investigated or analyzed.

This thesis is, therefore concerned with the three dimensional modelling and finite element analysis of the mandibular distortion.

The main aim of this thesis is to create a nucleus, for the computational biomechanics analysis of maxillofacial problems. The style has been aimed to satisfy both the dental/medical scientists and biomechanical engineers. Most of the fundamental anatomic and biomechanics data, assumed not known to either side, has been included in simplified form hopefully to generate further research, into one of the most challenging anatomic systems of the body.

CHAPTER II

ANATOMY OF THE MANDIBLE.

The masticatory system is the functional unit of the body primarily responsible for chewing, swallowing, and speaking. Components also play a major role in tasting and breathing. The masticatory system is a complex and highly refined unit. It is made up primarily of bones, joints, muscles, ligaments and teeth. Movement is regulated by an intricate neurologic controlling mechanism. Each movement is co-ordinated to maximize function while minimizing the energy used and damage to any structure. Precise movement of the mandible by the musculature is required to move the teeth efficiently across each other during function. A sound understanding of its functional anatomy and biomechanics is essential to the study of occlusion. There is no way of studying function without the knowledge of structure and no understanding of structure without knowledge of function, even at the molecular level. An appreciation of an integration of morphology and physiology of the oral structures is necessary for those who are concerned with the form and functions of the masticatory system.

2.1 The Skull.

The skull is the most complex bony structure in the body because it encloses the brain which is irregular in shape, houses the organs of special senses, and encloses the openings into the digestive and respiratory tracts.

Five views of the exterior of the skull are used in an atomical descriptions; each is spoken of as a norma.

i) Anterior aspect (Norma Frontalis):
The anterior aspect of the skull comprises the anterior part of the calvaria (brain case) superiorly and the skeleton of the face inferiorly. Notable features are: the forehead, the prominences of the cheek (zygomatic bones), the anterior nasal appertures and the paired maxillae (upper jaw) containing maxillary teeth and the mandible (lower jaw) containing the mandibular (lower) teeth.

The anterior aspect of the skull may be divided into five areas: frontal, maxillary, nasal, orbital and mandibular (Fig. 1).

In the anatomical position, the skull is oriented so that the inferior margins of the orbits (eye sockets) and the superior margins of the external acoustic meatus (auditory canals) are horizontal. This is called the orbitomeatal plane (Frankfurt plane).

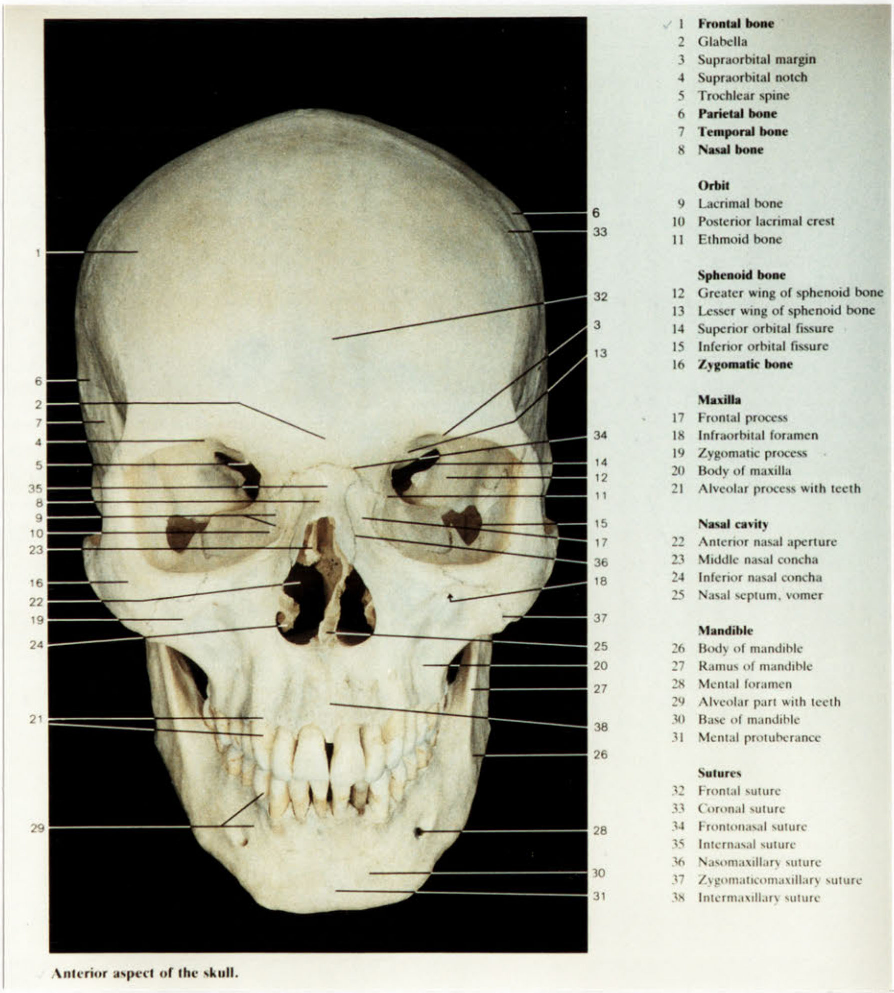


Figure 1 Photograph of the anterior aspect (norma frontalis) of an adult skull (Rohen and Yokochi, 1983).

ii) Posterior aspect (Norma Occipitalis):

The posterior aspect of the skull (Fig. 2), noticeably convex, is mainly formed by the parietal and occipital bones which meet the mastoid parts of the temporal bones laterally. The most prominent feature of the posterior aspect of the skull is the rounded posterior pole, called the occiput.

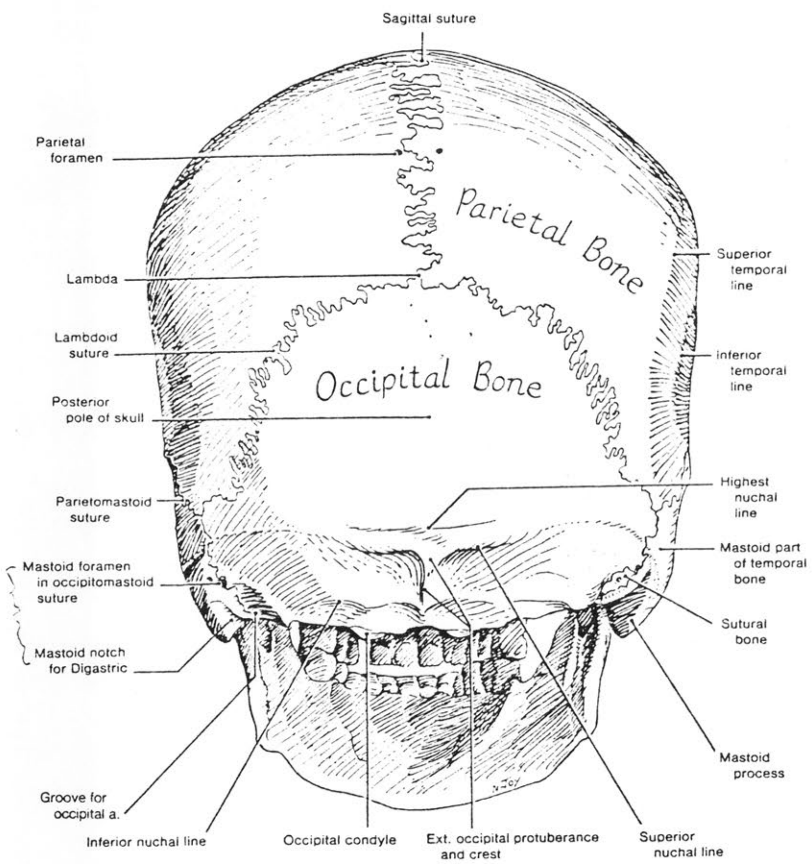


Figure 2. Drawing of the posterior aspect (norma occipitalis) of an adult skull, (Moore, 1982).

iii) Superior Aspect (norma Verticalis)

The superior aspect or "top of the skull" is rounded or void in many people and is broadened posteriorly by pariental eminences or tubers (Fig. 3).

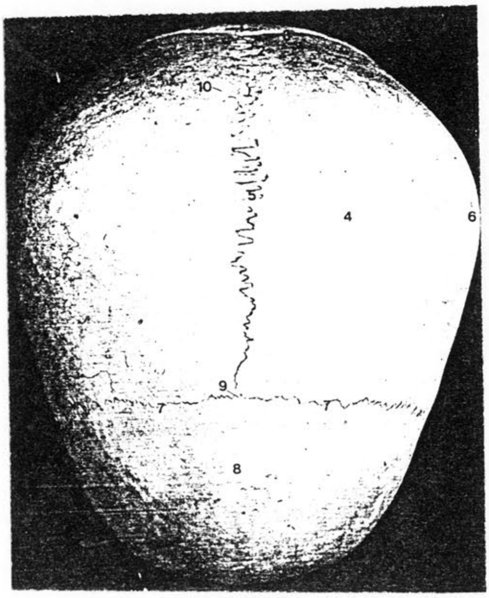


Figure 3. Photograph of the superior aspect (normal verticalis) of an adult skull, showing the sagittal and coronal sutures of the calvaria, (McMinn and Hutchings, 1977).

iv) Inferior Aspect (Norma Basalis).

The external surface of the base of the skull, with the mandible removed (Fig. 4), shows the inferior surface of the maxillae anteriorly, with the bony palete and maxillary teeth, and the zygomatic arches curving posteriorly on each side. The zygomatic arch is formed by the temporal process of the zygomatic bone and the zygomatic process of the temporal bone (zygoma). Centrally the inferior surface of the skull is irregular owing to the many foramina, processes, and articulations. Laterally the base of the skull exhibits the temporal bones with their prominent mastoid and styloid processes.

The foramen magnum, one of the most conspicuous features of the base of the skull, is bordered anterolaterally by the occipital condyles, which articulate with the atlas or first cervical vertebra.

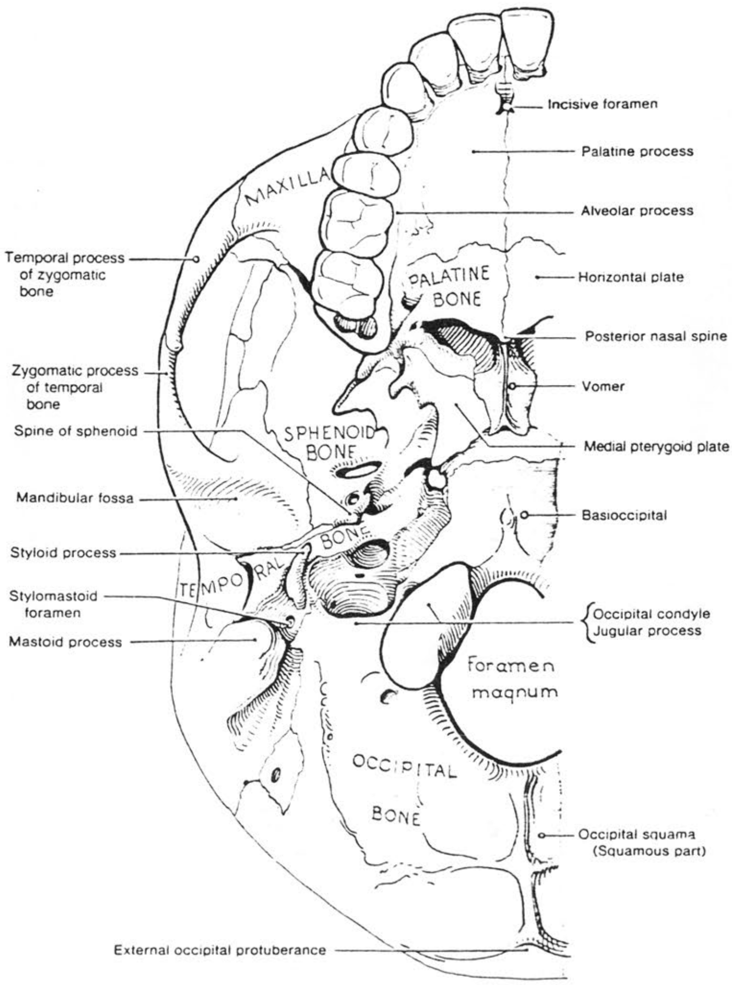


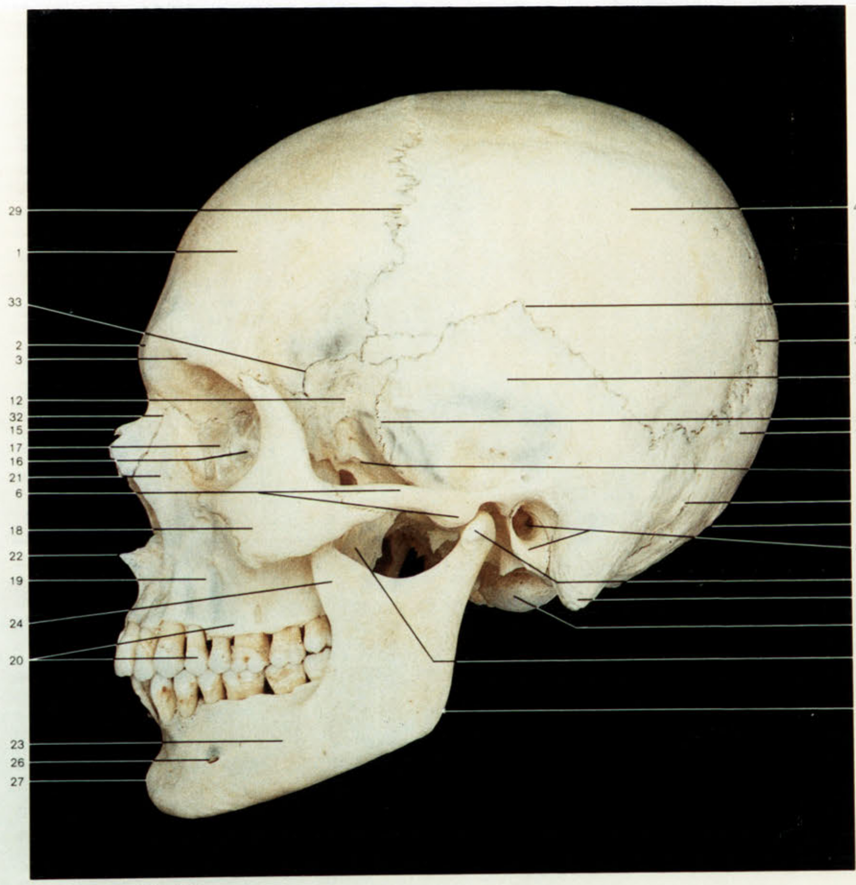
Figure 4. Drawing of slightly more than half of the external surface of the base (norma basalis) of an adult male skull. Note that the temporal process of the zygomatic bone unites with the zygomatic process of the temporal bone to form the zygaomatic arch (Moore, 1982).

v) Lateral Aspect (Norma Lateralis).

The lateral aspect of the skull (Fig. 5) includes parts of the temporal bone and the temporal and infratemporal fossae. The lateral aspect of the skull indicates clearly the division of the skull into the large ovoid cranial vault (brain case) and the smaller uneven facial skeleton. Obvious features of the lateral aspect of the skull are: the external acoustic meatus, zygomatic bone, zygomatic arch, mastoid process of the temporal bone, and mandible.

The mastoid process projects anteroinferiorly, medial to the lobule of the auricle, where it is readily palpable. The size of the mastoid process varies with the age and muscularity of the person. It is not present at birth and it is small during childhood. The mastoid process is part of the insertion of the sternocleidomastoid muscle.

The mental protuberance is an obvious feature in lateral views of the skull in most people. It consists of a triangular prominence of bone with its apex directly superiorly toward the incisive fossa of the mandible.



Lateral aspect of the skull.

1 **Frontal bone**
2 Glabella
3 Supraorbital margin
4 **Parietal bone**
5 **Temporal bone** (squamous part)
6 Zygomatic process, articular tubercle
7 Mastoid process
8 Tympanic part (tympanic plate) and external acoustic meatus
9 **Occipital bone** (squamous part)
10 External occipital protuberance
11 Occipital condyle
12 **Sphenoid bone,** greater wing of sphenoid bone
13 Infratemporal crest of sphenoid
14 Pterygoid process, lateral pterygoid plate
15 **Nasal bone**
16 **Ethmoid bone** (orbital part)
17 **Lacrimal bone**
18 **Zygomatic bone**
19 **Maxilla** (body)
20 Alveolar process and teeth
21 Frontal process
22 Anterior nasal spine
23 **Mandible** (body)
24 Coronoid process
25 Condyloid process
26 Mental foramen
27 Mental protuberance
28 Angle of the mandible

Sutures

29 Coronal suture
30 Lambdoid suture
31 Squamous suture
32 Nasomaxillary suture
33 Frontosphenoid suture
34 Sphenosquamosal suture
35 Occipitomastoid suture

Figure 5. Photograph of a lateral view (normal lateralis) of an adult skull, naming the bones and various outline features, (Rohen and Yokochi, 1982).

The skull is composed of many bones which are closely fitted together. Except for the mandible and the ossicles of the middle ear, the bones of the adult skull are joined by rigid structures of synchondroses. In effect, the cranium of a mature adult is essentially a single complex bone.

Although the adult skull is rigid, the bones of the cranium of infants and children grow as individual bones and undergo remodelling. Furthermore, relationships among the various bones are constantly changing during infancy and childhood.

2.2 The Mandible

The mandible is the largest and strongest bone of the face. Mandibular teeth project superiorly from their sockets in the alveolar process of the mandible.

The mandible consists of two parts: a horizontal part called the body, and two vertical oblong parts, the rami (Fig. 6). The right and left parts of the body are fused anteriorly at the symphysis menti to form a U shaped bone. The halves of the newborn mandible articulate at the symphysis menti. Fusion of this joint occurs in the second year.

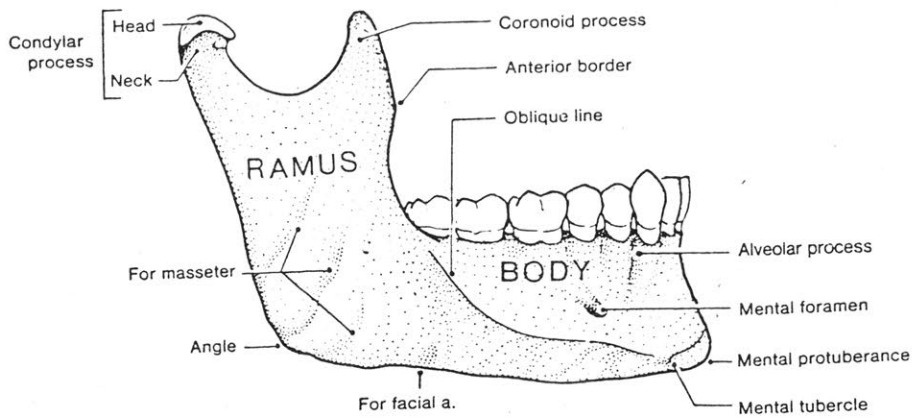


Figure 6: Drawing of the lateral aspect of the right half of an adult mandible, (Moore, 1982).

Each ramus ascends almost vertically from the posterior aspect of the body. The superior part of the ramus has two processes: a posterior condylar process with a head and neck, and a sharp anterior coronoid process. The condylar process is separated from the coronoid process by the mandibular notch, which forms the concave superior border of the mandible. The rami and the body meet posteriorly at the angle of the mandible.

In the anatomical position, the rami of the mandible are almost vertical, except during infancy and in edentulous persons.

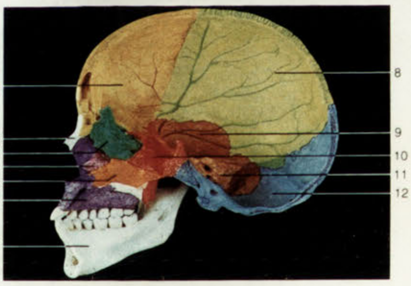
Viewed from the superior aspect, the mandible is horse shoe-shaped, whereas each half is L shaped when viewed from the side.

On the internal aspect of the ramus of the mandible, there is a large mandibular foramen (Fig. 7). It is the oblong entrance to the mandibular canal which transmits the inferior alveolar vessels and nerve to the roots of the mandibular teeth. Branches of these vessels and the mental nerve merge from the mandibular canal at the mental foramen.

Anterior to the mandibular foramen there is a thin, tongue-like projection (spur) of bone, called the lingula of the mandible, that somewhat overlaps and guards the superoanterior border of the foramen like a tongue or shield. The sphenomandibular ligament attaches to the lingula.

Running inferiorly and slightly anteriorly on the internal surface of the mandible from the mandibular foramen is a small mylohyoid groove (sulcus), indicating the course taken by the mylohyoid nerve and vessels.

The internal surface of the mandible is divided into two areas by the mylohyoid line, which commences posterior to the third molar tooth.



Median section through the skull. Internal aspect of the right half.

1 Hypophysial fossa
2 Anterior clinoid process
3 Frontal bone
4 Ethmoidal air cells
5 Sphenoidal sinus
6 Superior concha
7 Middle concha
8 Maxillary hiatus
9 Inferior concha
10 Inferior meatus
11 Anterior nasal spine and maxilla
12 Mental spine or genial tubercle
13 Groove for middle meningeal artery
14 Dorsum sellae
15 Internal acoustic meatus
16 Groove for sigmoid sinus
17 Hypoglossal canal
18 Occipital condyle
19 Condylar process
20 Lateral pterygoid plate
21 Medial pterygoid plate
22 Lingula of mandible
23 Mandibular foramen
24 Mylohyoid groove
25 Mylohyoid line
26 Submandibular fossa

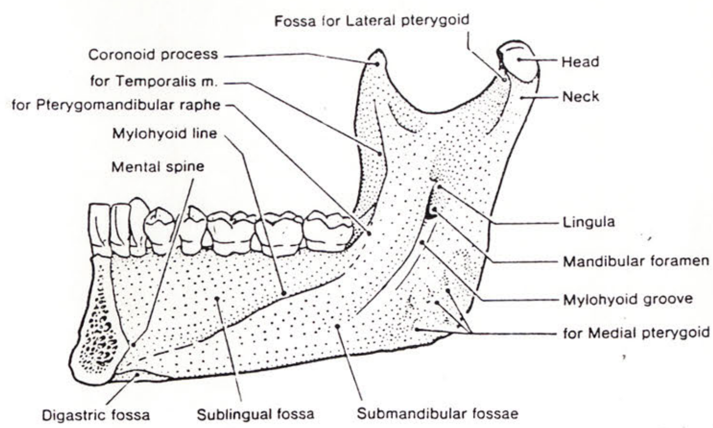


Figure 7 Photograph and a drawing of the internal aspect of the right half of an adult mandible (Rohen and Yokochi, 1983; Moore, 1982).

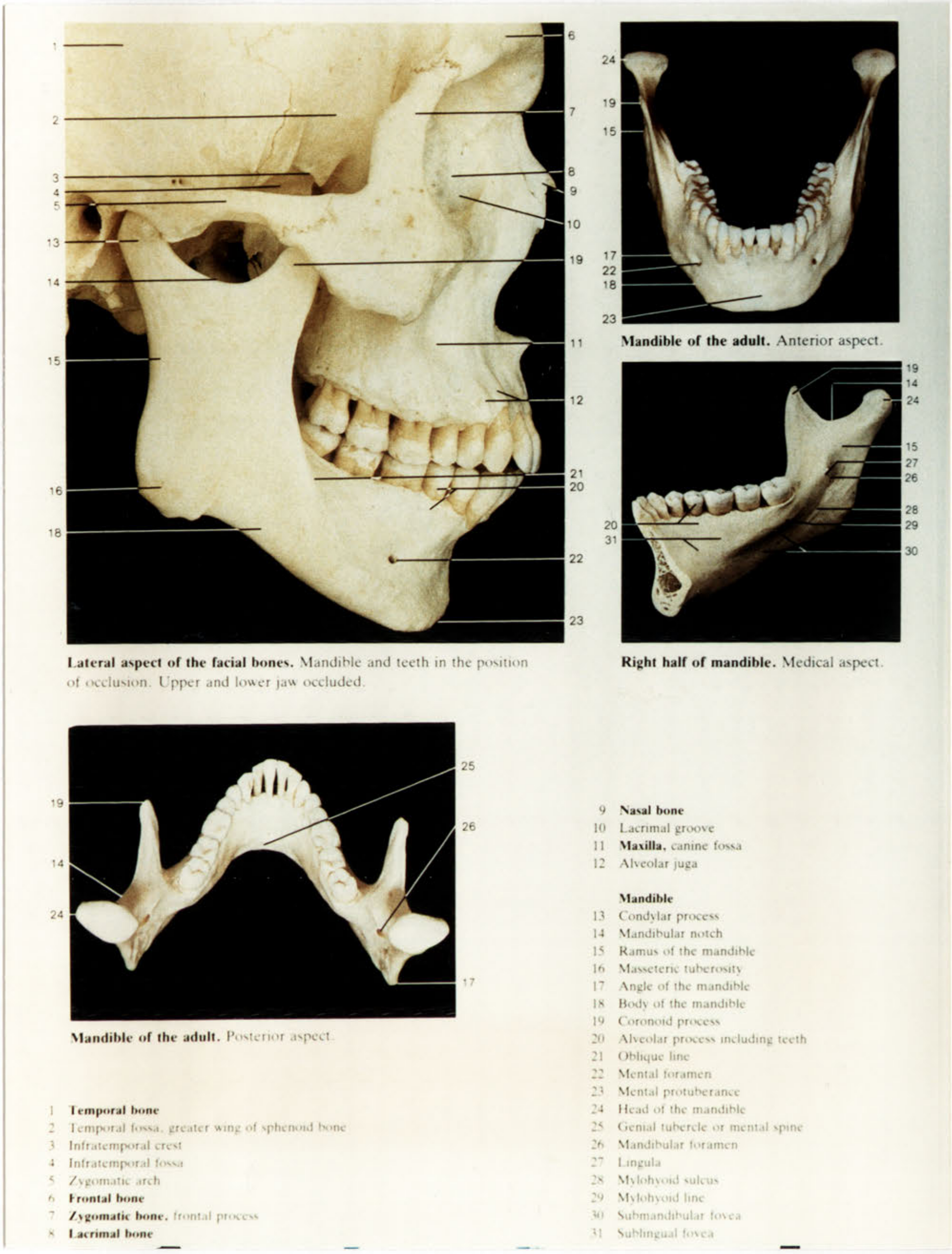
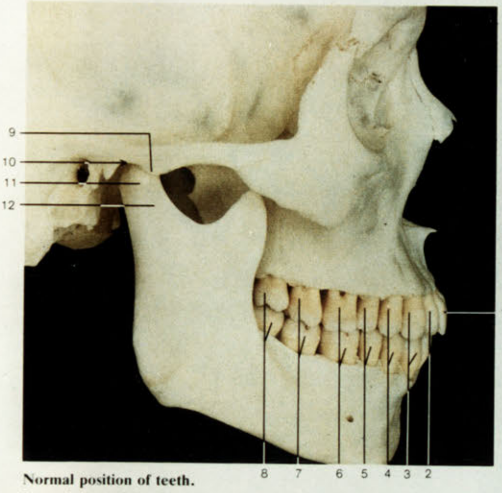
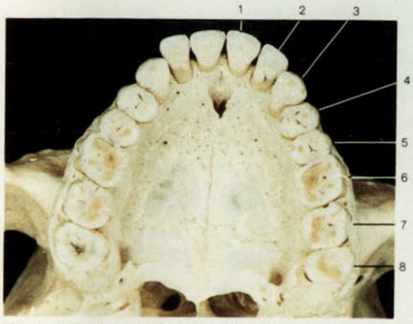


Figure 8 Photographs of the different aspects of the mandible (Rohen and Yokochi, 1983).

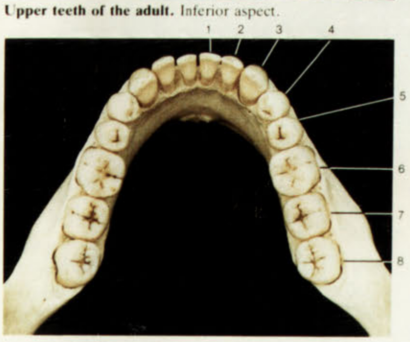


1 Central incisor
2 Lateral incisor
3 Canines
4 First premolars or bicuspids
5 Second premolars or bicuspids
6 First molars
7 Second molars
8 Third molars
9 Articular tubercle
10 Mandibular fossa
11 Head of mandible
12 Condylar process

Normal position of teeth.
Lateral view of the dentition in centric occlusion.



Upper teeth of the adult. Inferior aspect.



Lower teeth of the adult. Superior aspect.

Table of dentition

Primary dentition (Deciduous teeth)	Maxilla months post partum		Mandible months	
1. Central incisor	10.3		8.6	
2. Lateral incisor	12.2		14.4	
3. Cuspid incisor	19.5		20.1	
4. First molar	15.5		16.5	
5. Second molar	24.8		24.5	
Secondary dentition (Permanent teeth)	♂ years and months	♀	♂ years and months	♀
1. Central incisor	7/8	7/5	6/10	6/7
2. Lateral incisor	8/11	8/6	7/11	7/7
3. Cuspid incisor	12/2	11/7	11/12	10/3
4. First premolar	10/5	10/1	11/3	10/8
5. Second premolar	11/4	11/1	12/0	11/7
6. First molar	6/7	6/6	6/5	6/3
7. Second molar	12/9	12/5	12/3	11/9

Eruption of deciduous and permanent teeth
(after C. Röse according to A. Kröncke)

Figure 9 Photographs of the mandible maxilla and the position of teeth (Rohen and Yokochi 1983).

CHAPTER III

FUNCTIONAL ANATOMY

The mandibular condyle articulates at the base of the cranium with the squamous portion of the temporal bone. This portion of the temporal bone is made up of a concave mandibular fossa, in which the condyle is situated and which has also been called the articular or glenoid fossa (Fig. 10, 11).

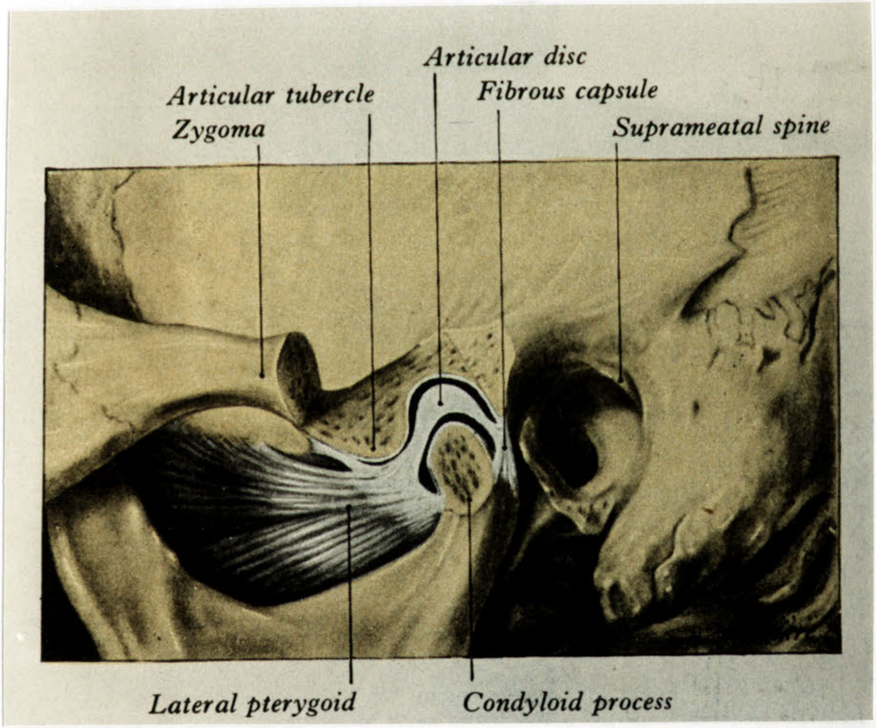


Figure 10 A sagittal section through the left temporomandibular joint (Gray's Anatomy, 1982).

Immediately anterior to the fossa is a convex bony prominence called the articular eminence. The degree of convexity of the articular eminence is highly variable but important since the steepness of this surface dictates the pathway of the condyle when the mandible is positioned anteriorly.

3.1 Temporomandibular Joint:

The area where craniomandibular articulation occurs is called the temporomandibular joint (TMJ). The temporomandibular or craniomandibular articulation is the articulation between the mandible and the cranium. It is a highly specialized synovial joint and is distinguished from most other such articulations by the avascular fibrous tissue that covers the bony articulating surfaces. This avascular fibrous tissue may contain a variable number of cartilage cells. This is unlike the usual condition where synovial joint articulating surfaces are covered by hyaline cartilage.

The temporomandibular joint is a diarthrodial joint because an articular disc is interposed between the temporal bone and the mandible, thus dividing the articular space into an upper and lower compartments.

Gliding movements occur in the upper compartment (arthrodial joint), while the lower compartment functions primarily as a synovial joint, (ginglymoid joint). Therefore, the temporomandibular joint is often classified as a hinge joint with a movable socket. (ginglymoarthrodial joint)

The temporomandibular joint is formed by the mandibular condyle fitting into the mandibular fossa of the temporal bone. Separating these two bones from direct articulation is the articular disc. The articular disc is composed of dense fibrous connective tissue devoid of any blood vessels or nerve fibres. In the sagittal plane it can be devided into three regions according to thickness. The central area is the thinnest and is called the intermediate zone. Both anterior and posterior to the intermediate zone the disc becomes considerably thicker. The posterior border is generally slightly thicker than the anterior border. In the normal joint the articular surface of the condyle is located on the intermediate zone of the disc, bordered by the thicker anterior and posterior regions, (Fig. 11).

From an anterior view the disc is generally thicker medially than laterally, which corresponds to the increased space between the condyle and the articular fossa toward the medial of the joint (Fig. 11).

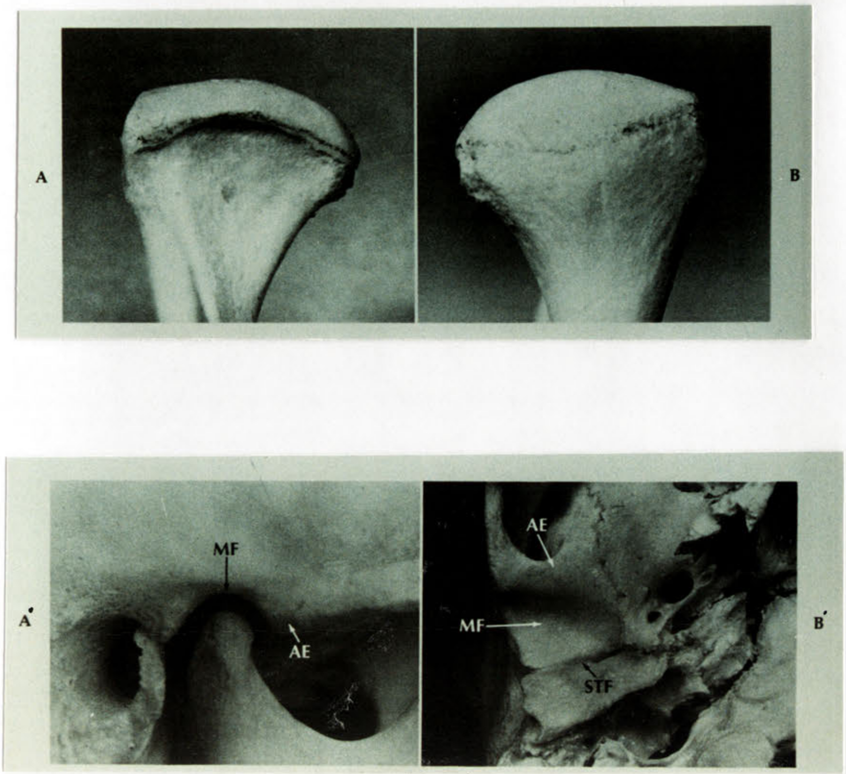


Figure 11 Articular disc, fossa and condyle, anterior view (A), posterior view (B), lateral view (A'), and inferior view (B'). MF mandibular fossa, AE articular eminence, STF, squamotympanic fissure (Okeson, 1985).

The precise shape of the disc is determined by the morphology of the condyle and mandibular fossa. During movement the disc is somewhat flexible and can adapt to the functional demands of the articular surfaces. The disc maintains its morphology unless destructive forces or structural changes occur in the joint. If these changes occur, the morphology of the disc can be irreversibly altered and a pathologic condition ensue.

The articular disc is attached to the capsular ligaments, which divides the joint into two distinct cavities. The upper (superior) cavity is bordered by the mandibular fossa and the superior surface of the disc. The lower (inferior) cavity is bordered by the mandibular condyle and the interior surface of the disc.

The internal surfaces of the cavities are surrounded by specialized endothelial cells that form a synovial lining. This lining, along with a specialized synovial fringe located at the anterior border of the retrodiscal tissues (Fig. 12), produces synovial fluid, which fills both joint cavities.

This synovial fluid serves two purposes. Since the articular surfaces of the joint are nonvascular, the synovial fluid acts as a medium for providing metabolic requirements to these tissues. There is free and rapid exchange between the vessels of the capsule, the synovial fluid, and the articular tissues. The synovial fluid also serves as a lubricant between articular surfaces during function.

Synovial fluid lubricates the articular surfaces by way of two mechanisms. The first is called boundary lubrication, which occurs when the joint is moved and the synovial fluid is forced from one area of the cavity into another. The synovial fluid located in the border regions is forced upon the articular surface, thus providing lubrication. A second lubricating mechanism is called weeping lubrication. This refers to the ability of the articular surfaces to absorb a small amount of synovial fluid. Thus, when articular surfaces are placed under compressive forces this small amount of synovial fluid is released, lubricating the tissues. Weeping lubrication helps eliminate friction in the compressed but not moving joint. Only a small amount of friction is eliminated as a result of weeping lubrication, and prolonged compressive forces to the articular surfaces might exhaust this supply.

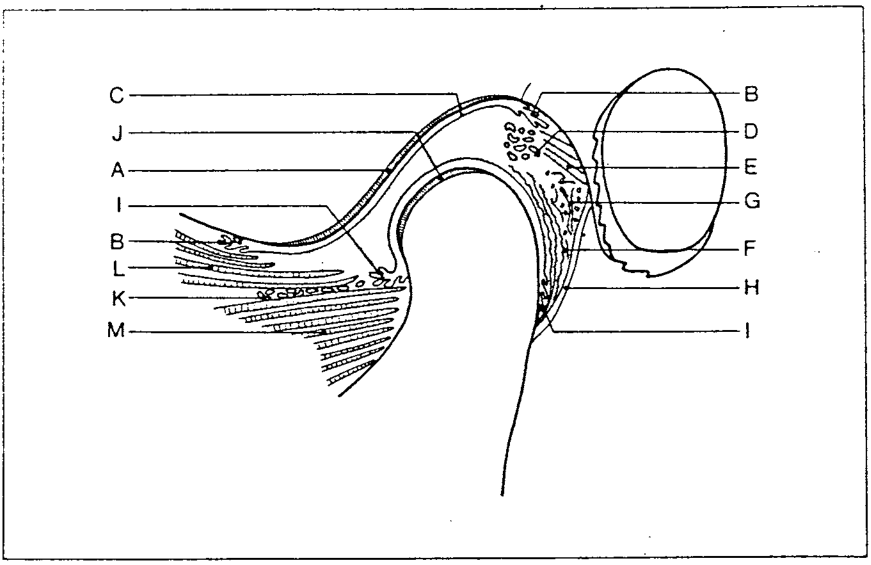


Figure 12 Temporomandibular joint (Sagittal view); A, articular surface, B, synovial membrane, C, superior cavity, D, vascular knee, E, superior stratum, F, inferior stratum, G, loose areolar connective tissue, H, posterior capule, I, inferior cavity synovial membrane, J, articular surface of condyle, K, blood vessels, L, superior belly of lateral pterygoid, M, inferior belly of lateral pterygoid muscle. (from Bell, W. D. Clinical management of temporomandibular disorders, Chicago 1982).

3.2 Temporomandibular Joint Ligaments:

As in any joint system, ligaments play an important role in protecting the structures. The ligaments of the joint are made up of collagenous connective tissues, which do not stretch. They do not enter actively in joint function but instead act as passive restraining devices to limit and restrict joint movement. There are three functional and two accessory ligaments that support the TMJ:

i) The collateral (discal) ligaments,
ii) the capsular ligament,
iii) the temporomandibular ligament,
iv) the sphenomandibular ligament, and
v) the stylomandibular ligament.

3.2.1 The Collateral Ligaments:

The discal or collateral ligaments attach the medial and lateral borders of the articular disc to the poles of the condyle. There are two collateral ligaments. The medial discal ligament attaches the medial edge of the disc to the medial pole of the condyle. The lateral discal ligament attaches the lateral edge of the disc to the lateral pole of the condyle (Fig. 13).

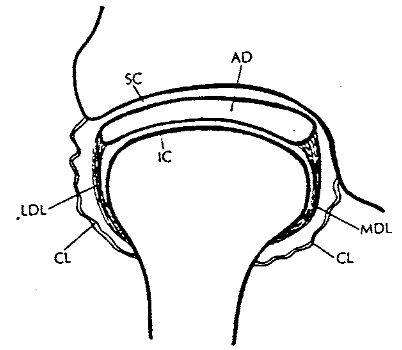


Figure 13 Temporomandibular Joint (anterior view), AD, articular disc, CL, capsular ligament, LDL, lateral discal ligament, MDL, medial discal ligament, SC, superior joint cavity, IC, inferior joint cavity. (from Mahan, P.E., and Kreitziger, K.L: In Alling, C.C., and Mahan, P.E.: Facial pain, 1977).

These ligaments are responsible for dividing the joint mediolaterally into the superior and inferior joint cavities. The discal ligaments are true ligaments, composed of collagenous connective tissue fibres, and therefore do not stretch. They function to restrict movement of the disc away from the condyle. In other words, they cause the disc to move passively with the condyle as it glides anteriorly and posteriorly. The attachments of the discal ligaments permit the disc to be rotated anteriorly and posteriorly on the articular surface of the condyle. These ligaments thus are responsible for hinging type movement of the TMJ, which occurs between the condyle and the articular disc.

The discal ligaments have a vascular supply and are innervated. Their innervation provides information regarding joint position and movements. Strain on these ligaments produces pain.

3.2.2 Capsular Ligament:

The entire TMJ is surrounded and encompassed by the capsular ligament (Fig. 14). The fibres of the capsular ligament are attached superiorly to the temporal bone along the borders of the articular surfaces of the mandibular fossa and articular eminence. Inferiorly the fibres of the capsular ligament attach to the neck of the condyle.

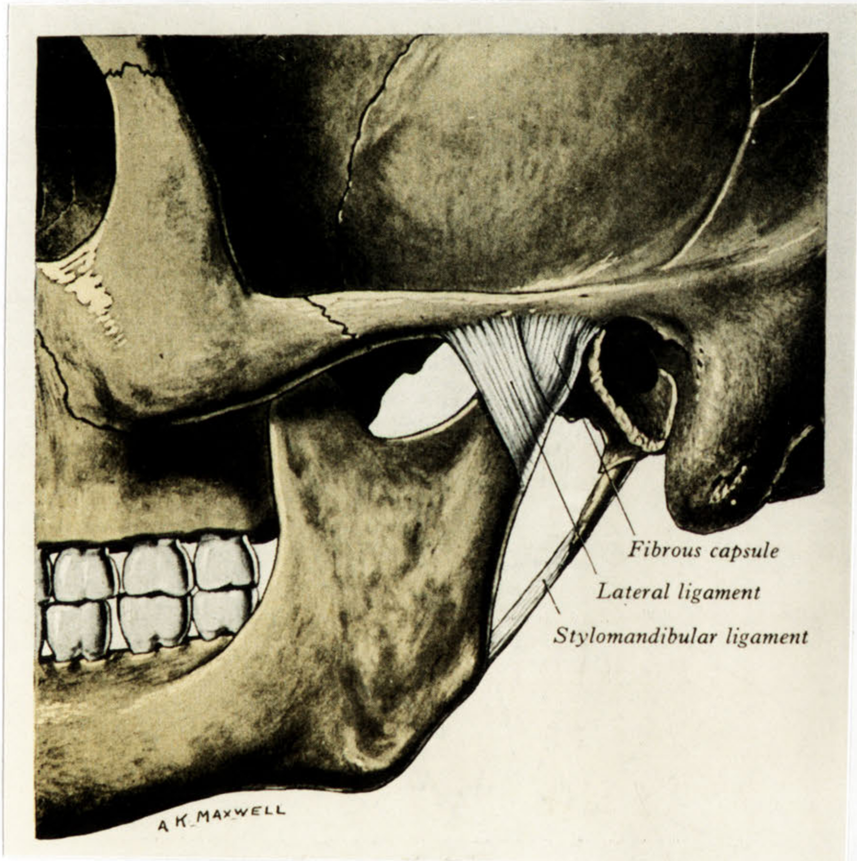


Figure 14 Capsular ligament (lateral view)

The capsular ligament acts to resist any medial, lateral or inferior forces that tend to separate or dislocate the articular surfaces. A significant function of the capsular ligament is to encompass the joint thus retaining the synovial fluid. The capsular ligament is well innervated and provides proprioceptive feed back regarding position and movement of the joint.

3.2.3 Temporomandibular Ligament:

The lateral aspect of the capsular ligament is reinforced by strong tight fibres that make up the lateral ligament or the temperomandibular ligament. The TM ligament B composed of two parts, an outer oblique portion and an inner horizontal portion (Fig. 15).

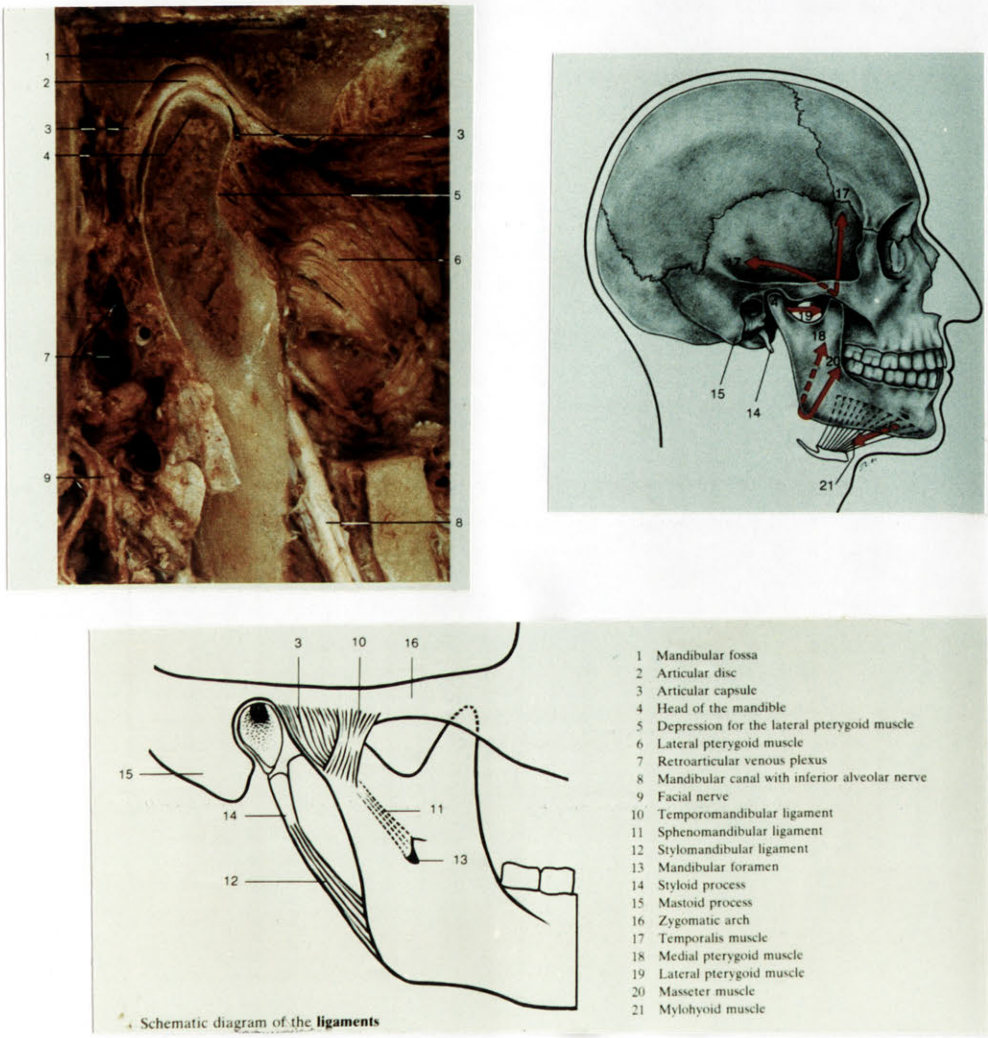


Figure 15 Photograph and drawings of Temporomandibular joint, a) sagittal section through the temporomandibular joint, b) effect of the mastication muscles on the temporomandibular joint, c) schematic diagram of the ligaments (Rohen and Yokochi, 1983).

The outer portion extends from the outer surface of the articular tubercle and zygomatic process poster oinferiorly to the outer surface of the condylar neck. The inner horizontal portion extends from the outer surface of the articular tubercle and zygomatic process posteriorly and horizontally to the lateral pole of the condyle and posterior part of the articular disc.

3.2.4. Sphenomandibular Ligament:

The spenomandibular ligament is one of two accessory ligaments of the TMJ (Fig. 16). It inserts on the mandible at the mandibular lingula, which is located along the anterior and superior border of the mandibular foramen. It also inserts along the lower border of the groove of the mandibular neck. In most individuals, the sphenomandibular ligament is a thin layer of connective tissue with indistinct anterior and posterior borders. It does not have any significant limiting effects on mandibular movement. It has been suggested by Moss (1959) that this ligament protects the neural and vascular structures passing through the mandibular foramen from tensile stress during mouth opening. This hypothesis assumes that the axis of mandibular rotation is always located within the region of the mandibular foramen.

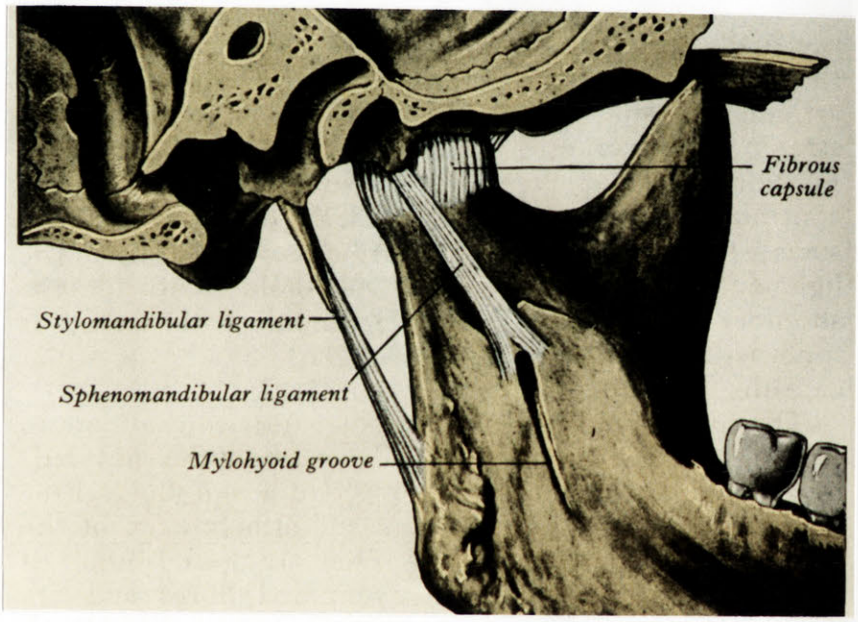


Figure 16 The mandible, TMJ and the accessory ligaments (medial aspect) (Gray's Anatomy, 1982).

3.2.5. The Stylomandibular Ligament:

The stylomandibular ligament is a reinforced part of a fascial lamella that extends from the styloid process and stylohyoid ligament to the region of the mandibular angle. Part of its fibres are attached to the mandible itself, but the majority continue into the fascia on the medial surface of the medial pterygoid muscle. The upper border of the stylomandibular ligament is often thickened considerably. The ligament is loose when the jaw is closed and is tense when the mandible is protruded. This ligament might function in limiting excessive protrusive movements. The stylomandibular ligament is in its most relaxed state during maximal opening.

3.3 The Musculature of the Masticatory Mechanism:

The muscles involved in jaw movements include the muscles of mastication and also other muscles namely suprahyoid and infrahyoid groups:

i) Muscles of mastication:
Masseter;
Temporalis;
Medial pterygoid;
Lateral pterygoid.

ii) Suprahyoid Muscles:
Digastric;
Stylohyoid;
Mylohyoid;
Geniohyoid.

iii) Infrahyoid Muscles:
Sternohyoid;
Thyrohyoid;
Sternothyroid;
Omohyoid.

3.3.1 Muscles of Mastication:

The masticatory muscles, especially the elevators, act in directions specifically designed to transmit the majority of the masticatory force in a direction at right angles to the dental arches. The muscles also tend to stabilize the occlusion in contact so that no rocking or tipping occurs. This specific force direction and stabilization is the result of the combined action of the temporalis, masseter, medial and lateral pterygoid muscles.

The muscles of mastication, in conjunction with the tongue, facial, palatal, infra and suprahyoid muscles, work as a group. Nevertheless, knowledge·of the theoretical action of the individual mandibular muscles is necessary in analyzing their co-ordinate function during movements of the mandible.

3.3.1.1 Masseter Muscle:

The masseter is a quadrilateral muscle, consisting of three superimposed layers which blend with one another in the anterior region. The superficial layer is the largest and arises as a thick aponeurosis from the zygomatic process of the maxilla, and from the anterior two thirds of the lower border of the zygomatic arch (Fig. 17, 18). Its fibres pass downwards and backwards to insert into the ramus and angle of the mandible. The middle layer arises from the deep surface of the anterior two-thirds of the zygomatic arch and from the lower border of the posterior third and is inserted into the middle of the ramus of the mandible. (Fig.29, 30)

The deep layer arises from the deep surface of the zygomatic arch and inserts into the upper of the ramus of the mandible and to the coronoid process. The fibres of the deep layer run vertically downwards and slightly forwards. Immediately in front of the TMJ, the deep portion is not covered by the superficial portion and can be seen as a triangular muscle field, the fibres of which run almost exactly downward at an angle of about 30 to 40 degrees to the fibres of the superficial muscle splat. The deep part of the masseter muscle is inseparably fused with the most superficial fibres of the temporalis muscle.

The oxygen consumption of the masseter muscle greatly exceeds that of the limb muscles, indicating some difference in their energetic functional process (Kawamura and Takata 1961). The number of muscle fibres per motor unit in the masseter is high which indicates that the masseter muscle is a powerful muscle. Sicher (1960), states that "muscle composed of fibres arranged at an angle to the axis of the muscle will consist of relatively more and shorter fibres and will therefore, by primarily muscles of great power". He says that the masseter muscle has this specific structure.

The action of the muscle is that of a powerful elevator which exerts pressure at a right angle to the molar region of the dental arch. The deep portion of the muscle has a retracting component in its action. Therefore the muscle acts in elevating and to a small degree retruding the mandible. The fibres of the deep portion that originate immediately in front of the articular eminence is thought to stabilize the condyle against the articular eminence during biting (Hylander, 1975).

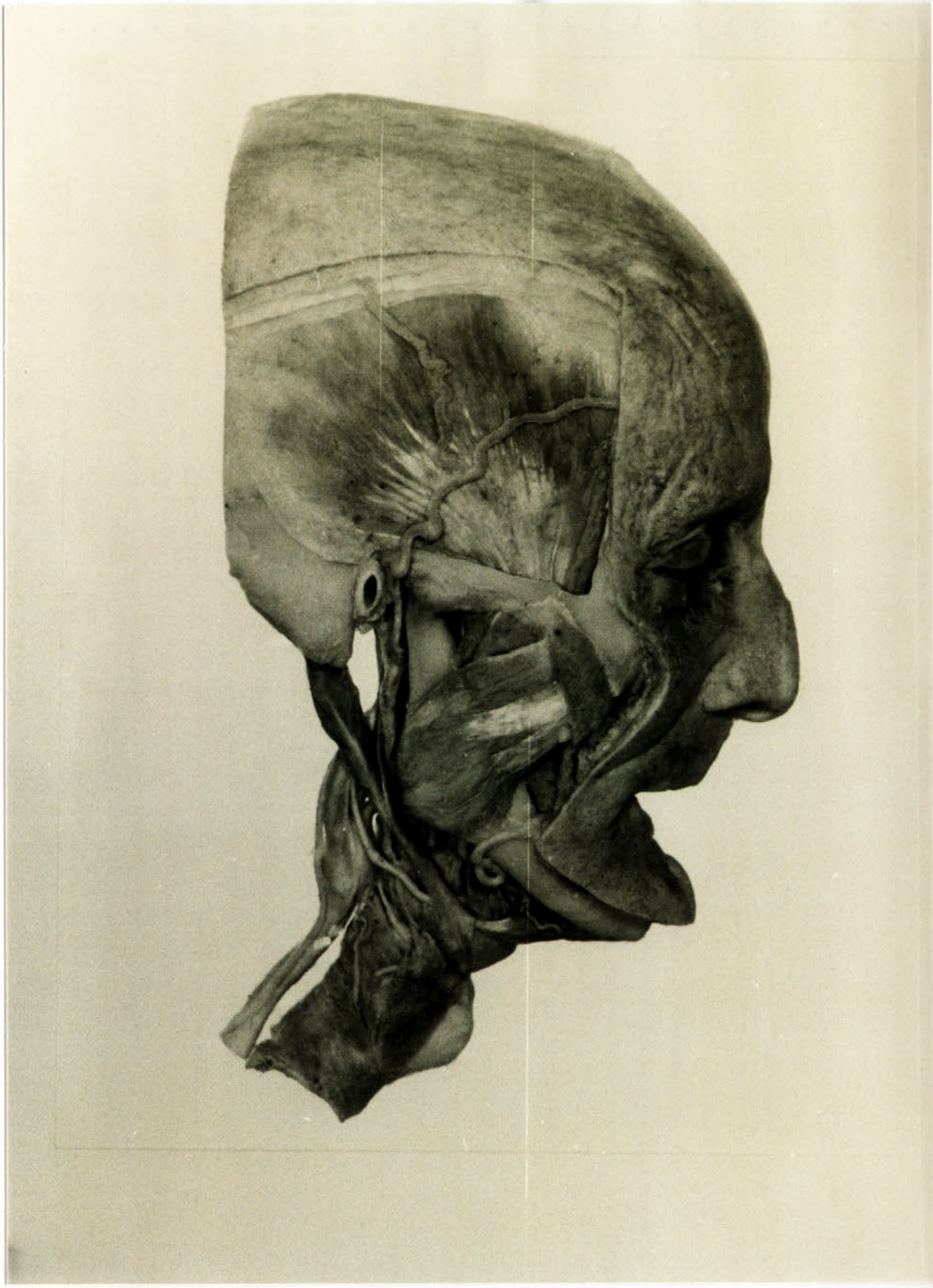


Figure 17 Mandible, TMJ, and muscles of mastication.

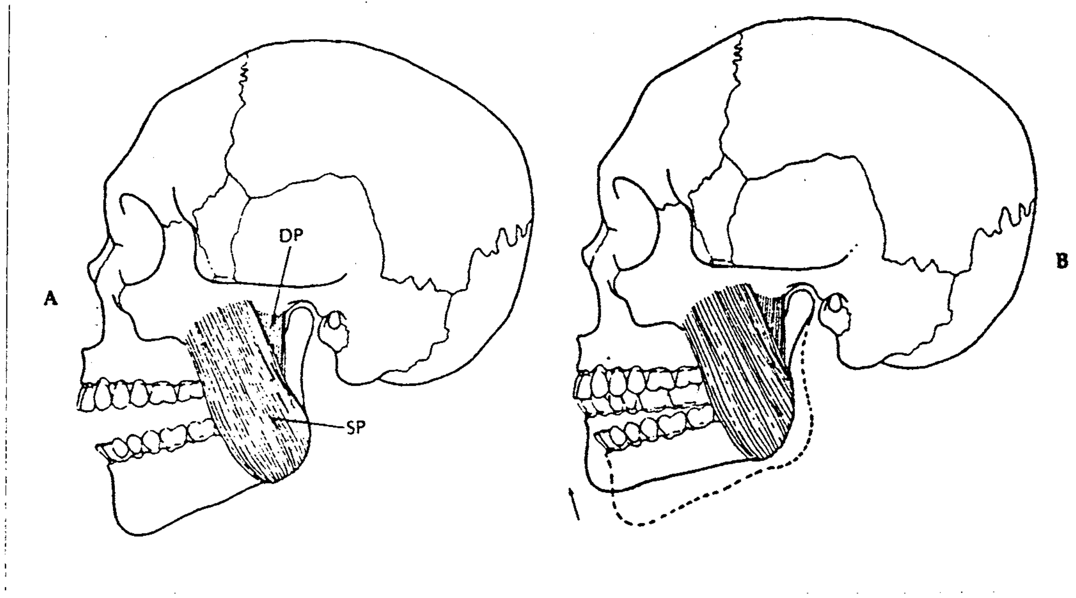


Figure 18 Masseter muscle (A). SP, superficial portion, DP, deep portion. Function of the masseter muscle during elevation of the mandible (B).

3.3.1.2 The Temporalis:

The temporalis is a large fan-shaped muscle that originates from the temporal fossa and the lateral surface of the skull. Its fibres come together as they extend downward between the zygomatic arch and the lateral surface of the skull to form a tendon that inserts on the coronoid process and anterior border of the ascending ramus (Fig.29, 30). It can be divided into three distinct areas according to fibre direction and alternate function. The anterior portion consists of fibres that are directed almost vertically. The middle portion contains fibres that run obliquely across the lateral aspect of the skull. The posterior portion consists of fibres that are aligned almost horizontally, coming forward above the ear to join other temporalis fibres as they pass under the zygomatic arch (Fig. 19).

When the entire temporalis contracts, it elevates the mandible, and the teeth are brought into contact. If only portions contract, the mandible is moved according to the direction of those fibres that are activated. When the anterior portion contracts, the mandible is raised vertically. Contraction of the middle portion will elevate and retrude the mandible. Function of the posterior portion is somewhat controversial. Some authors suggest that contraction of this portion will retrude the mandible; others suggest that contraction of the posterior portion will cause elevation and only slight retrusion.

It really appears that the most posterior fibres should have a large mandible-retracting component because of their oblique direction downward and forward; however, these fibres are bent around the root of the zygomatic process and thus are oriented essentially in a vertical manner (Sicher and Du Brul,1975). Therefore, this portion of the temporalis muscle exerts a vertical force on the mandible. Since its fibres pass close to the articular eminence, it probably functions both as a stabilizer of the TMJ and an elevator of the mandible.

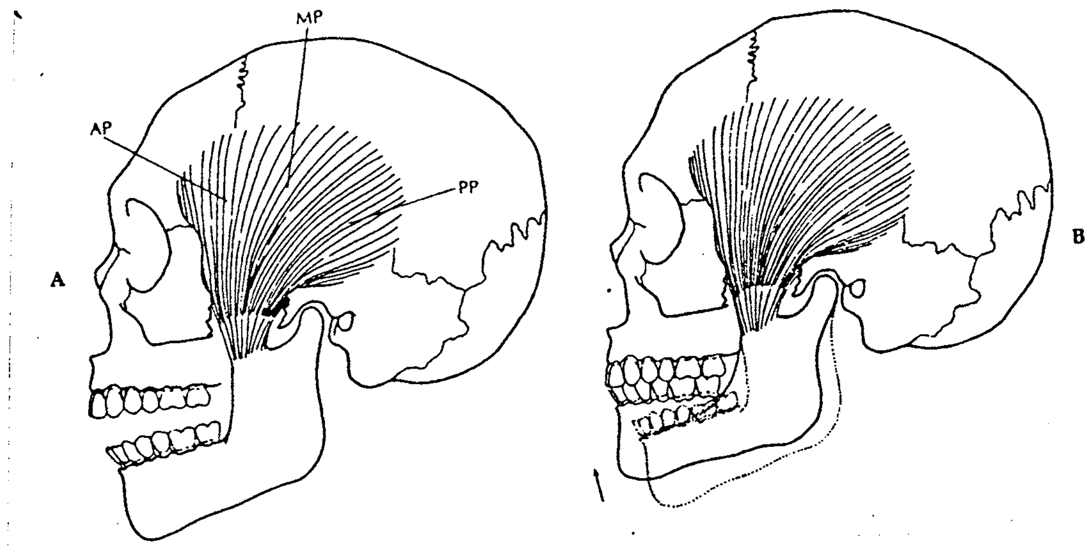


Figure 19 Temporalis muscle (A), AP, anterior portion, MP, middle portion, PP, posterior portion. Function of the temporalis muscle during elevation of the mandible(B).

3.3.1.3 The Medial (Internal) Pterygoid:

The medial pterygoid muscle, which is situated on the medial side of the mandibular ramus, is a counterpart of the masseter muscle, both anatomically and functionally. Although it is a powerful rectangular muscle, it is weaker than the masseter. Its main origin is in the pterygoid fossa. The innermost fibres arise by strong tendons, while others arise directly from the medial surface of the lateral pterygoid palate. The tendon covers the medial surface of the muscle at its origin and is as wide as the tensor veli palatini muscle with which it is in contact. The more anterior fibres arise from the outer and inferior surface of the pyramidal process of the palatine bone and from the adjacent parts of the maxillary tuberosity. These fibres are positioned lateral to the lateral pterygoid muscle. The remaining and largest portion of this muscle is positioned medial to the lateral pterygoid muscle.

The fibres of the medial pterygoid muscle run downward, backward and outward and are inserted along the medial surface of the mandibular angle. The field of insertion is approximately triangular and is located between the mandibular angle and the mylohyoid groove (Fig.20, 30, 31). The fibres of the medial pterygoid muscle often meet fibres of the masseter in tendinous raphe behind and below the mandibular angle. This muscular arrangement is ofter termed the pterygomasseteric sling.

The internal structure of the medial pterygoid muscle is a complicated alternation of fleshy and tendinous parts. The muscle fibres themselves, arising from one tendon and ending on another, are arranged at an angle to the general orientation of the muscle. This arrangement gives the muscle fibres a braided appearance and tends to increase the power of the muscle. The fibre orientation of the medial pterygoid muscle is similar to the superficial portion of the masseter muscle and therefore it is primarily an elevator of the mandible.

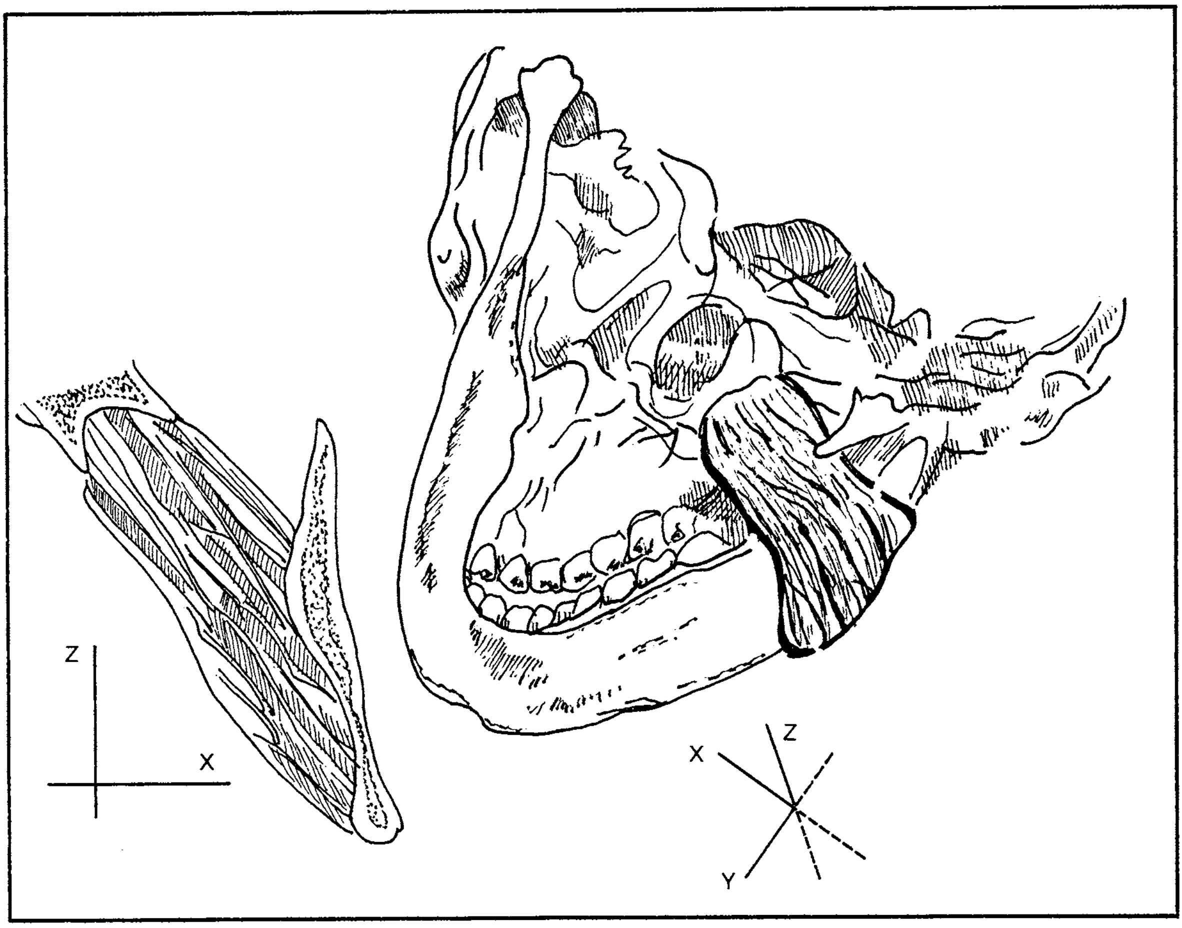


Figure 20 Medial Pterygoid (inferior and posterior aspects), (Hawthorne, 1969).

3.3.1.4 The Lateral (External) Pterygoid Muscle:

The lateral pterygoid muscle arises from two heads. The larger inferior head originates from the outer surface of the lateral pterygoid plate, and the smaller superior head originates from the infra temporal surface of the greater sphenoid wing medial to the infra temporal crest. The fibres of the upper head run almost horizontally backward and outward in close relation to the cranial base. The fibres of the lower head converge upward and outward and ascend in a progressively steep manner. The two heads, separated at their origins by a wide gap, supposedly fuse in front of the TMJ. (Sicher 1960) The uppermost and deepest fibres, which constitute the upper head, are attached to the articular capsule and thus to the anterior border of the articular disc. According to Sicher (1960), the majority of the fibres of the superior head and all of the fibres of the inferior head insert along a roughened fossa on the anterior surface of the mandibular neck. Honee (1972), however stated that the two heads do not fuse in front of the TMJ and that the superior head is only attached to the capsule and disc. Therefore it can be identified as two distinct and different muscles.

The lateral pterygoid muscle is thought to have three heads of insertion (Hawthorn 1969), while most of the anatomical evidence indicates that the lateral pterygoid muscle is made up of two functionally distinct parts, (McNamara 1973 and Lipke et al. 1977).

The inferior lateral pterygoid originates at the outer surface of the lateral pterygoid plate and extends backward, upward, and outward to its insertion primarily on the neck of the condyle (Fig. 21). When the right and left inferior lateral pterygoids contract simultaneously, the condyles are pulled down the articular eminences and the mandible is protruded. Unilateral contraction creates a mediotrusive movement of that condyle and causes a lateral movement of the mandible to the opposite side. When this muscle functions with the mandibular depressors, the mandible is lowered and the condyles glide forward and downward on the articular eminences.

The superior lateral pterygoid is considerably smaller than the inferior and originates at the infratemporal surface of the greater sphenoid wing, extending almost horizontally, backward, and outward to insert primarily on the articular capsule and disc and to a slight extent on the neck of the condyle. Whereas the inferior lateral pterygoid is active during opening, the superior remains inactive, becoming active only in conjunction with the elevator muscles. The superior lateral pterygoid is especially active during the power stroke and when the teeth are held together.

The power stroke refers to movements that involve closure of the mandible against resistance, such as in chewing.

The pull of both lateral pterygoids on the disc and condyle is in a significantly medial direction. As the condyle moves more forward, the medial angulation of the pull of these muscles becomes even greater. In the wide open mouth position the direction of the muscle pull is almost entirely medial.

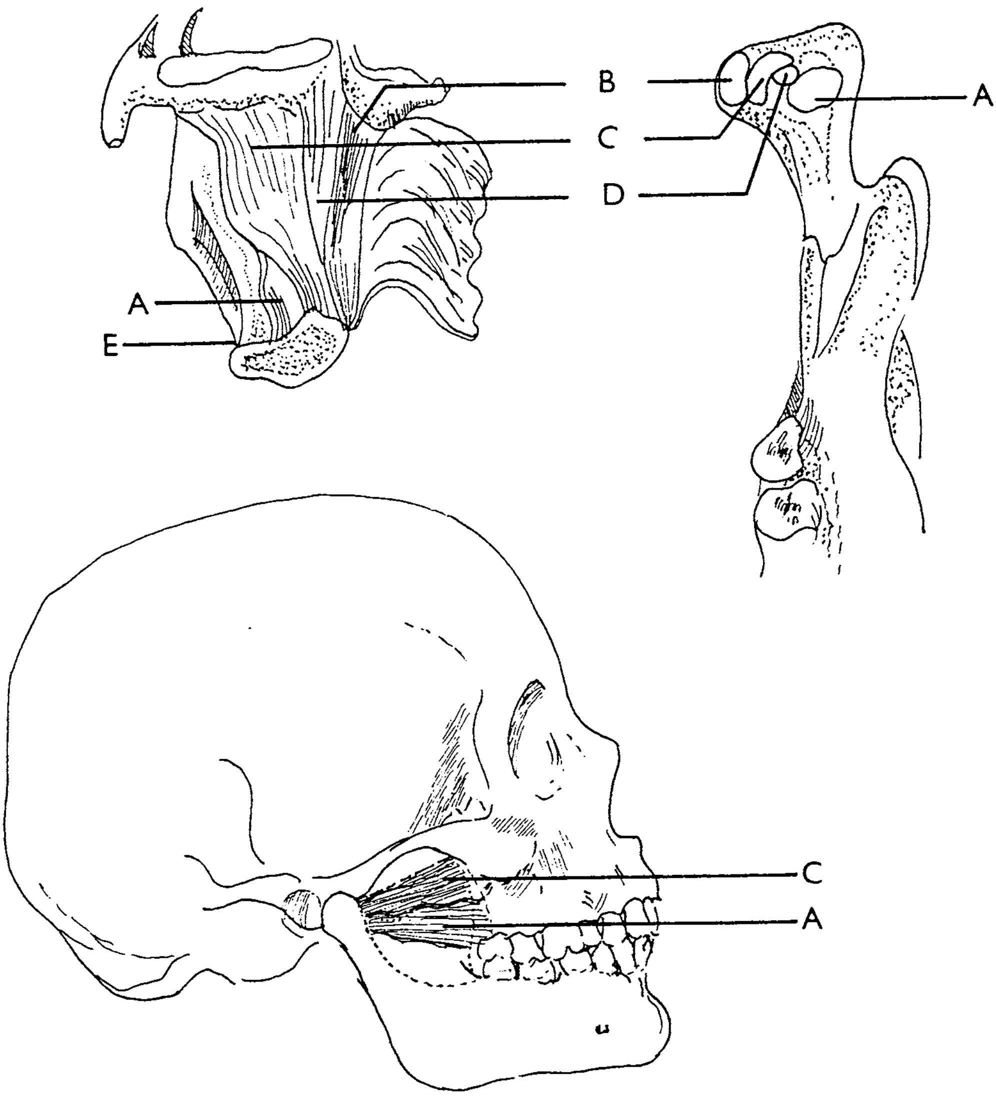


Figure 21 Lateral pterygoid muscle, A) inferior head, B) medial head, C) superficial slip of superior head, D) deep slip of superior head, E) ramus of mandible (redrawn from Hawthorne, 1969).

3.3.2 The Suprahyoid Muscles:

This muscle group is arranged between the skull, the mandible and the hyoid bone and functions either to elevate the hyoid bone and with it the larynx or to depress the mandible. Whether one or other movement is affected depends on the state of contraction of other associated muscles. If the mandible is fixed in position by the action of masseter, temporalis and medial pterygoid muscles in the tooth contact of swallowing, the suprahyoid group will elevate the hyoid bone and larynx, and facilitate swallowing. If on the other hand the infrahyoid muscles are contracted the hyoid bone is immobilized and the suprahyoid muscles that extend to the mandible will assist in depression and retraction of the lower jaw. The suprahyoid muscles are the digastric, the geniohyoid, the mylohyoid, and the stylohyoid (Fig. 22).

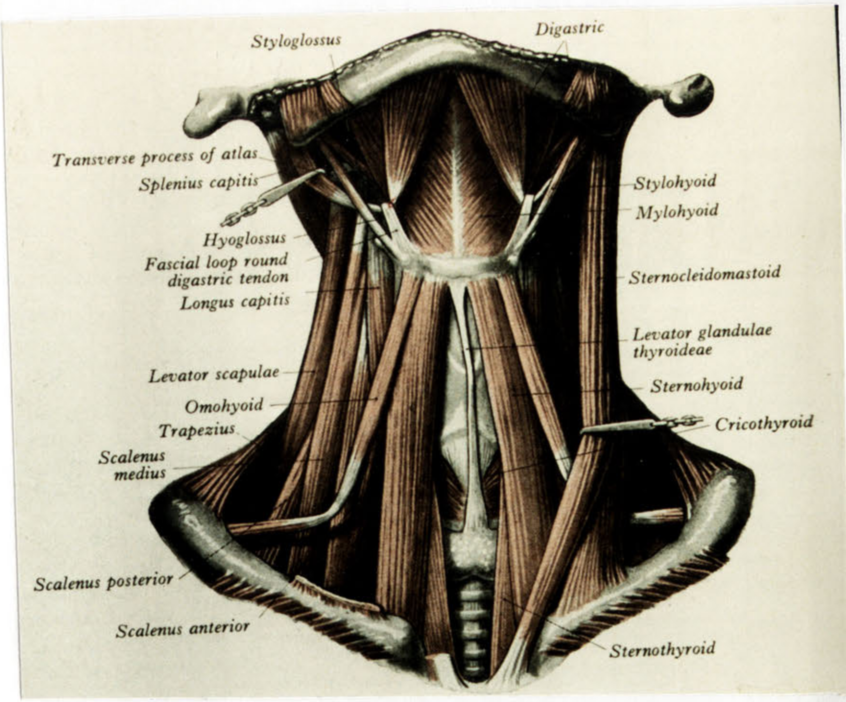


Figure 22 Muscles of the front of the neck (Gray's Anatomy, 1982).

3.3.2.1. The Digastric Muscle:

The digastric muscle consists of two fleshy parts, a posterior and an anterior belly, which are connected by a strong round tendon (Fig. 17, 23).

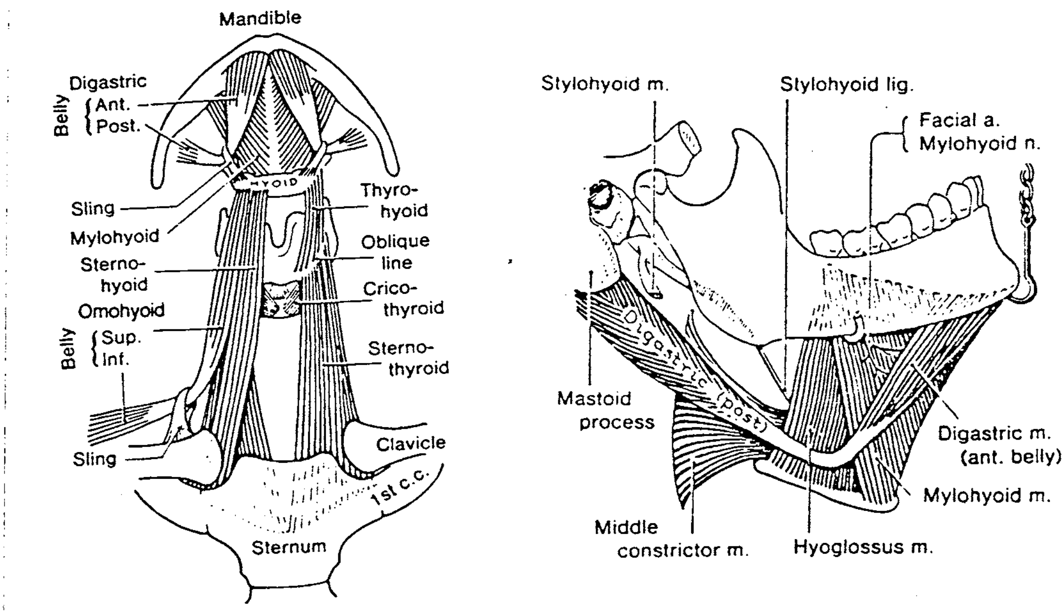


Figure 23 The digastric muscle (Moore , 1982).

The posterior belly arises from the mastoid notch medial to the mastoid process. The intermediate tendon is fixed by a fascial pulley to the hyoid bone, and the anterior belly finds its attachment in the digastric fossa of the mandible at its lower border close to the midline. The two bellies of the muscle form an obtuse angle that is maintained by the pulley of fascia, which fixes the tendon in its relation to the hyoid bone, but allows the tendon to slide in the fascial sling.

The posterior belly is much longer than the anterior belly, almost circular in cross section, and only slightly flattened in latero medial direction. Gradually tapering anteriorly, the posterior belly continues into the round intermediate tendon.

The anterior belly, arising from the intermediate tendon, is much shorter than the posterior belly. It consists, in most individuals, of a thicker lateral and a thinner medial part. Its insertion into the digastric fossa of the mandible is partly fleshy and partly tendinous. The transverse diameter of the anterior belly varies considerably. If this part of the muscle is broad, right and left muscles touch or almost touch each other at their insertion. If the anterior belly of the digastric muscle is narrow, there is a distance between right and left muscles, and the submental region is not triangular but irregularly quadrilateral.

The intermediate tendon is not directly attached to the hyoid bone but is fastened to it by strengthened fibres of the cervical fascia, which form a loop around the tendon. The tendon, therefore, can slide in this loop. The fibres of this pulley are attached to the greater horn and the lateral part of the body of the hyoid bone (Fig. 24)

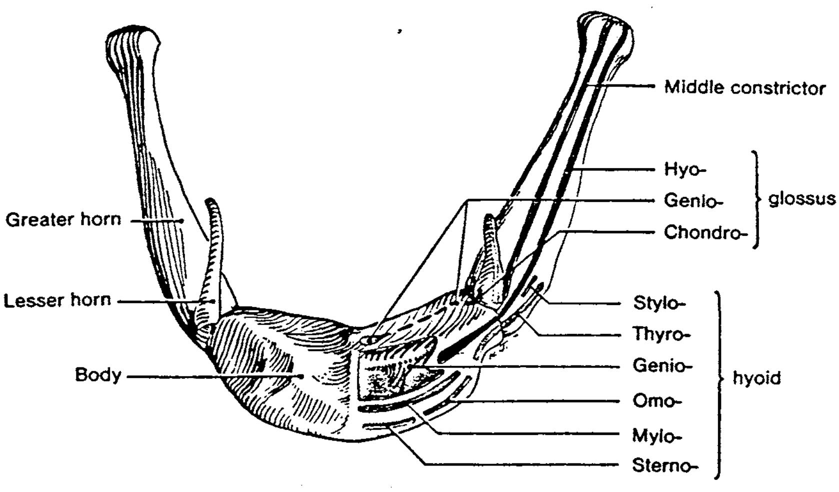


Figure 24 Drawing of the hyoid bone showing its parts and the sites of attachments of muscles (Moore, 1982).

The length of the fascial loop varies considerably; thus, the distance of the tendon from the hyoid bone and the angle between the posterior and anterior bellies of the digastric muscle also varies. The longer the loop the greater, therefore, the distance between hyoid bone and tendon, the more obtuse is the angle between the two bellies of the muscle.

It is usually stated that if the hyoid bone is fixed by the action of the stylohyoid and infrahyoid muscles, activity of the digastric muscle then pulls the mandible back and down. Thus, the digastric muscle is thought to function during retrusive and opening movements of the mandible. During opening, the digastric forms a force couple with the lower head of the lateral pterygoid muscle.

Moyers (1950) in his EMG studies found that the digastric muscles always act together and that they are secondary to the lateral pterygoids in mandibular depression, and coming into play especially in maximal depression.

Crompton et al.(1975) in his studies of hyoid movements during opening and closing of the jaws suggested that the hyoid bone is never completely fixed during these movements. Nevertheless, the digastric muscles appear to function as stated. The posterior belly is found to be active in swallowing and chewing actions.

3.3.2.2. The Geniohyoid Muscle:

The geniohyoid muscle (Fig. 25) arises above the anterior end of the mylohyoid line from the inner surface of the mandible, close to the midline and lateral to the mental spines, by a short and strong tendon. The muscle, in contact with that of the other side, proceeds straight posteriorly and slightly downward and is attached to the upper half of the hyoid body. Posteriorly the muscle gradually widens and assumes, in cross section, a triangular shape.

The geniohyoid muscle pulls the hyoid bone upward and forward, thus acting as a partial antagonist to stylohyoid, or it exerts a downward and backward pull on the mandible, depending on the fixation by other muscles, of either mandible or hyoid bone.

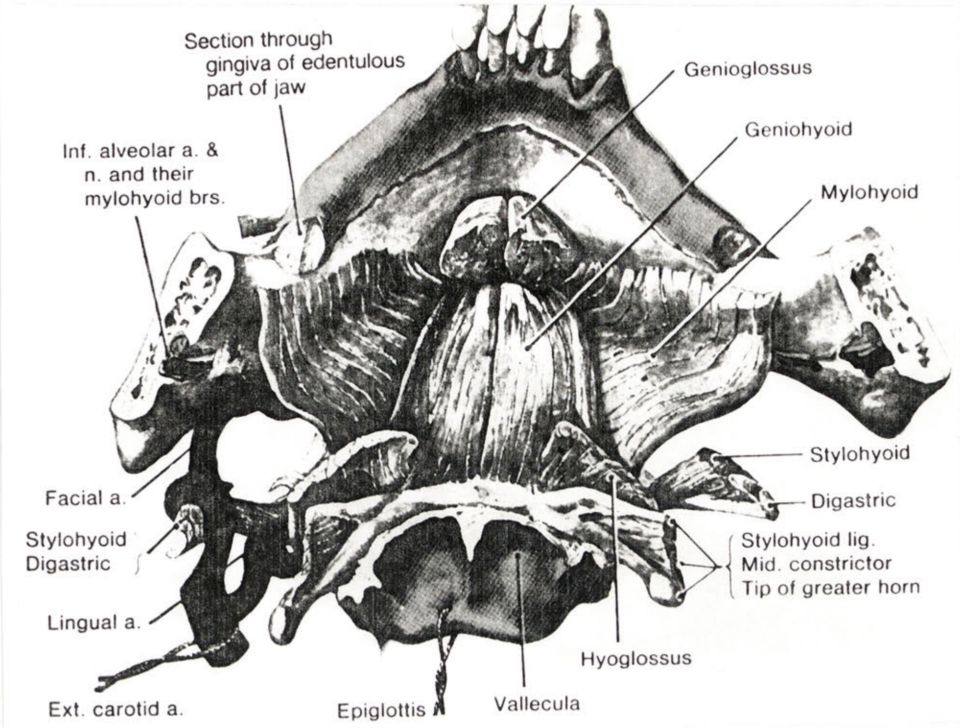


Figure 25 Geniohyoid and mylohyoid muscles in a drawing of a dissection of the muscles of the floor of the mouth (Moore, 1982).

3.3.2.3. The Mylohyoid Muscle:

The mylohyoid muscle form, anatomically and functionally, the floor of the oral cavity. The right and left muscles are united in the midline between mandible and hyoid bone by a tendinous strip, the mylohyoid raphe (Fig. 26).

The muscle arises from the mylohyoid line on the finer surface of the mandible. Its most posterior fibres take their origin from the region of the alveolus of the lower third molar. The origin of the anterior fibres deviates more toward the lower border of the mandible. The posterior fibres of the coarsely bundled muscle run steeply downward, medially, and slightly forward, and are attached to the body of the hyoid bone; the majority of the fibres, however, join those of the contralateral muscle in the mylohyoid raphe. The muscle plate is considerably thicker in its posterior part.

Only the posterior bundles of the mylohyoid muscle, running almost vertically from the mandible to the hyoid bone, is believed to have a slight, almost negligible, influence on these bones; if the hyoid bone is held in place, they help in depressing the mandible.

The anterior larger but thinner part of the mylohyoid muscle has no action on the lower jaw or the hyoid bone.

In contracting, the downward convex plate of this portion, consisting of the right and left fibres joined in the median raphe, flattens its curvature, and thus the floor of the oral cavity is elevated and with it the tongue that rest on it. The mylohyoid muscle is, therefore, tought to be primarily an elevator of the tongue.

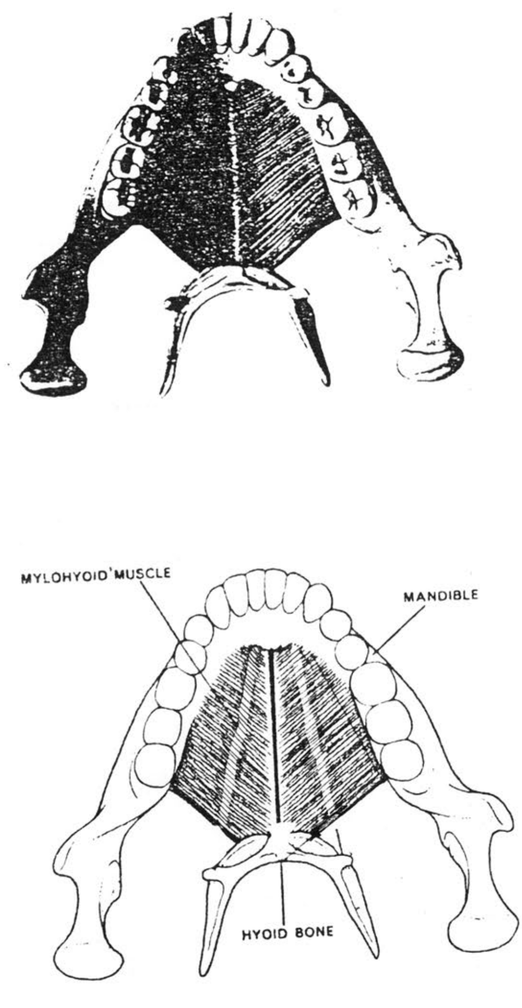


Figure 26 Mylohyoid muscle, superior view.

3.3.2.4 Stylohyoid Muscle:

The stylohyoid muscle arises from the lateral and inferior surface of the styloid process (Fig. 27). It is a thin round muscle that converges with the posterior belly of the digastric muscle anteriorly and inferiorly. It lies first superior and medial to the digastric muscle and then close to its upper border. Where the stylohyoid muscle reaches the intermediate tendon of the digastric muscle, it splits in most individuals into two slips that enclose the tendon of the digastric muscle and insert onto the greater horn of the hyoid bone where it joins the body.

The function of the stylohoid muscle is the elevation and retraction of the hyoid bone. This elongates the floor of the mouth. With the other supra and infra-hyoid muscles it can fix the hyoid bone as a basis for the action of tongue muscles attached to the bone. Its precise role in masticatory and vocal movements have not been satisfactorily analysed.

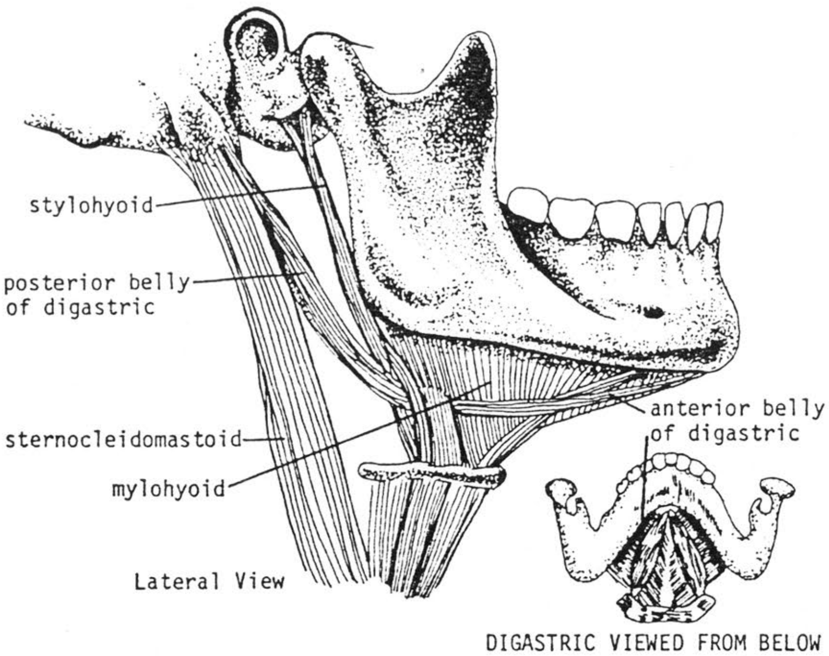


Figure 27 Stylohyoid muscle (lateral view).

3.3.3 Infrahyoid Muscles:

The infrahyoid muscles extend between the hyoid bone above the sternum, clavicle, and scapula below. The two superficial muscles, sternohyoid and omohyoid, directly connect the shoulder girdle and the sternum with the hyoid bone; the deep layer connects the sternum to the thyroid cartilage, and the omohyoid connects the scapula to the hyoid bone. Thus, the deep layer is divided into two muscles, the sternothyroid and the thyrohyoid (Fig. 28).

The function of the infrahyoid muscles is twofold; they may either depress the hyoid bone (Crompton et al. 1975) and, with it, the larynx, or together with the stylohyoid, fix the hyoid bone in its position, anchoring it to the trunk. The hyoid bone then is made the fixed point from which the suprahyoids, with the exception of the stylohyoid muscle, can act on the mandible. Although the larynx generally moves with the hyoid bone, the attachment of the deep layer of the infrahyoids to the thyroid cartilage allows for some independent movement of the larynx and hyoid bone.

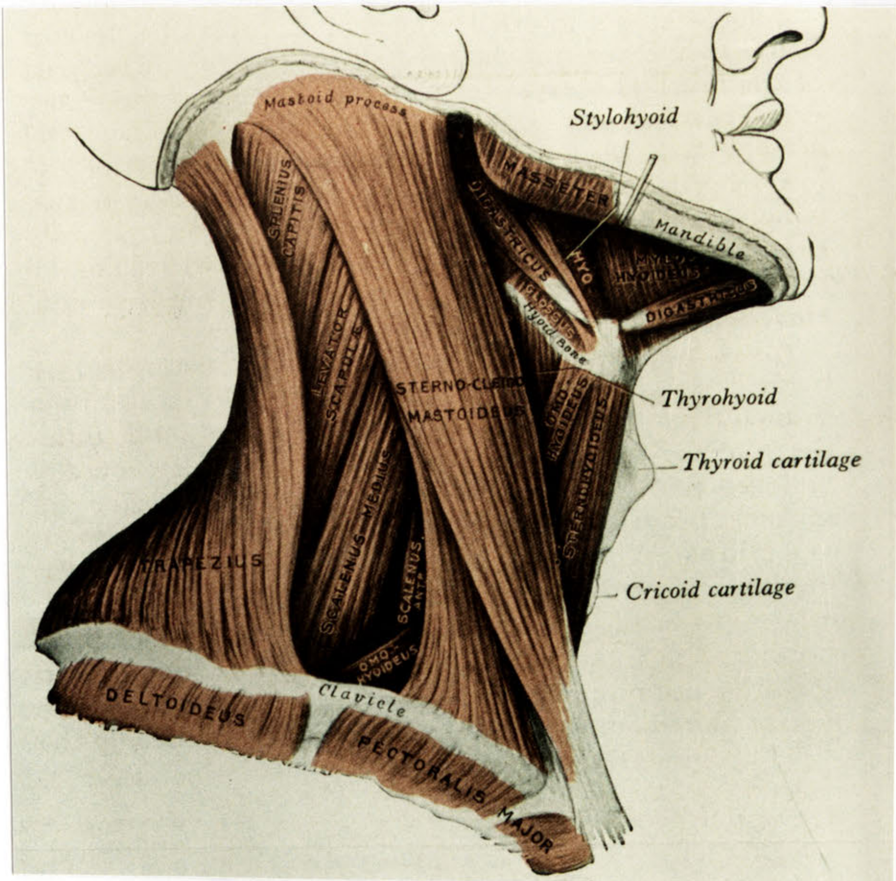
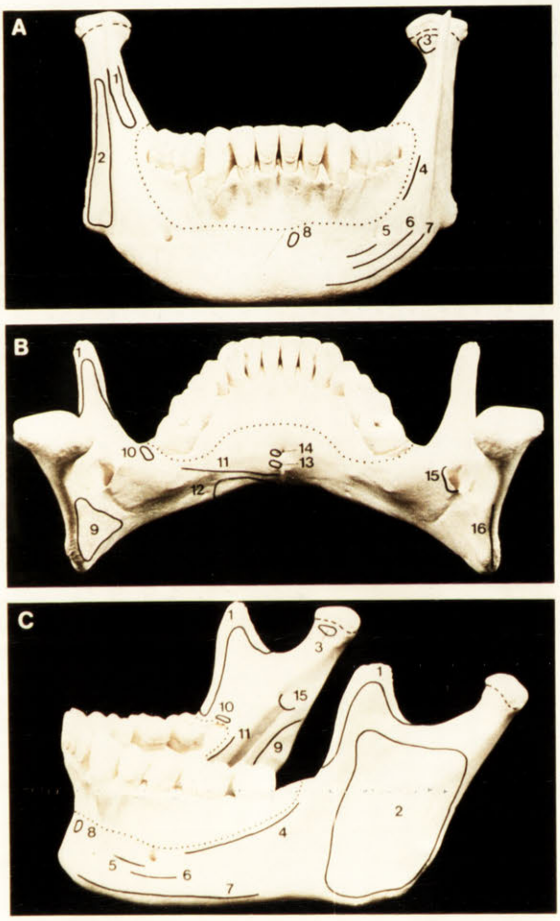


Figure 28 Infrahyoid muscles and the muscles of the neck (lateral aspect) (Gray's Anatomy, 1982).

The skull, from the right. Muscle attachments

1 Occipital part of occipitofrontalis
2 Sternocleidomastoid
3 Temporalis
4 Masseter
5 Zygomaticus major
6 Zygomaticus minor
7 Orbicularis oculi
8 Procerus
9 Levator labii superioris alaeque nasi
10 Levator labii superioris
11 Nasalis
12 Levator anguli oris
13 Buccinator
14 Depressor labii inferioris
15 Depressor anguli oris
16 Platysma

Figure 29 Insertion of the muscles (McMinn and Hutchings, 1977).



The mandible, A from the front, B from behind and above, C from the left and front. Muscle attachments
(Capsule attachment, interrupted line; the dotted line indicates the limit of attachment of the oral mucous membrane)

1 Temporalis
2 Masseter
3 Lateral pterygoid
4 Buccinator
5 Depressor labii inferioris
6 Depressor anguli oris
7 Platysma
8 Mentalis
9 Medial pterygoid
10 Pterygomandibular raphe and superior constrictor
11 Mylohyoid
12 Anterior belly of digastric
13 Geniohyoid
14 Genioglossus
15 Sphenomandibular ligament
16 Stylomandibular ligament

● The buccinator is attached to the alveolar processes of the maxilla and mandible opposite the three molar teeth. (Note that in this mandible the third molar has not yet erupted).

Figure 30 Insertion of the muscles (McMinn and Hutchings, 1977).

Inferior surface of the base of the skull.
Muscle attachments
(Capsule attachment, interrupted line)

1 Masseter
2 Upper head of lateral pterygoid
3 Deep head of medial pterygoid
4 Superior constrictor
5 Tensor veli palatini
6 Palatopharyngeus and musculus uvulae
7 Levator veli palatini
8 Pharyngeal raphe
9 Longus capitis
10 Rectus capitis anterior
11 Rectus capitis lateralis
12 Styloglossus
13 Stylohyoid
14 Stylopharyngeus
15 Posterior belly of digastric
16 Longissimus capitis
17 Splenius capitis
18 Sternocleidomastoid
19 Occipital part of occipitofrontalis
20 Trapezius
21 Semispinalis capitis
22 Superior oblique
23 Rectus capitis posterior minor
24 Rectus capitis posterior major

Figure 31 Insertions from the inferior aspect (McMinn and Hutchings, 1977).

CHAPTER IV

FUNCTIONAL ASPECTS OF THE MANDIBLE

The study of mandibular function is not limited only to the muscles of mastication. Other major muscles, such as the sternocleidomastoid and the posterior neck muscles, play major roles in stabilizing the skull and enabling controlled movements of the mandible to be performed. There exists a finely tuned dynamic balance of all the head and neck muscles, and this must be appreciated for an understanding of the physiology of mandibular movement to occur.

Movements of the mandible may occur as opening (depressive), closing (elevation), forward (protrusive), backwards (retrusive) or side to side (lateral) movements either alone or in combination with each other. Some very extensive studies have been carried out on human subjects in an attempt to define the movements of the mandible during chewing and to determine the role of the individual muscles in the execution of those movements. These experiments present enormous technical problems if the apparatus used to gather the information is not to interfere too much with the normal working of the system. To obtain a complete record of movements of the mandible during mastication, it would be necessary to monitor changes in its position in three planes with respect to the remainder of the skull.

The range of movements of the mandibular condyle were described in classical papers at the turn of the century by Ulrich (1896) and Bennett (1908). Ulrich observed the movement of a ball overlying the condyle which was attached by a wire to the lower teeth. Bennett used a light in a similar way, with a second one fixed to some other part of the mandible. These lights were projected with a lens onto a wall, where their movements were plotted on a piece of paper. With this method, movements of the whole mandible could be plotted and its instantaneous axis of rotation determined.

These very simple experiments together showed that there was no fixed axis of rotation for the mandible and that normal movements always involved both a simultaneous hinge and a sliding movement of the condyle. Even the small movement from rest to occlusion always involved a small posterior movement as well as a rotation of the condyle. With maximum opening the condyles move forward beyond the lowest part of the articular eminence, and further than they move during maximum protrusion, with the teeth in contact. When the mandible is moved to one side, the condyle on that side moves laterally (Bennett movement) and backward, and if the lateral movement of the jaw is accompanied by opening, then the condyle on that side moves downward too. It was also shown that a posterior movement of the condyles from centric occlusion can be produced. These observations nearly 90 years later still form the basis of present knowledge on the potential movements of the temporomandibular joint.

Similar experiments using tracing devices fixed at one end to the mandible and the other to a tracing plate was carried out by Campion (1905), Gysi (1921), Posselt (1952), Shanahan and Leff (1966).

Lindblom (1960) pointed out that it is nearly impossible with these methods to avoid a certain degree of pressure being applied to the jaws during registration of movement.

Thouren (1914) used a film camera with 16 exposures per second and thus obtained a moving picture of mandibular movements on a moving plate through a narrow slot using small indicators attached to the outside of the lower central incisors.

Cineradiography both direct and indirect has also been applied to the study of mandibular movements by Hildebrand (1931), Jankelson (1953) and Kydd (1958).

Several authors have used the cephalostat which restricts the subject's head movements and maxilla remains in a constant position with respect to the camera (Broadbent 1931), or modifications of it when taking radiographs of the temporamandibular joints and jaws and deducing mandibular movements therefrom [Bjork (1947); Posselt (1952); Nevarkari (1956); Brown (1965)].

To understand the functional movements of the mandible, detailed specific informations on the muscle activities has to be obtained. The simplest way of obtaining evidence of muscle activity in man is to attach electrodes to the skin and record the small electrical potentials set up by action currents flowing in the underlying muscles. Although this method of recording electromyograms from skin has the advantage that it is simple, it suffers some very serious limitations. The most obvious is that it cannot be used to record activity from muscles such as lateral pterygoid which are not immediately beneath the skin; the further the muscle fibres are from the electrodes, the smaller the recorded potentials tend to be. Its major limitation stems from the fact that the record obtained cannot always be attributed with certainty to one particular muscle. This is a very serious limitation when one attempts to record from the anterior belly of digastric. Two electrodes placed over this muscle, just medial to the lower border of the mandible, will pick up activity from mylohyoid, geniohyoid, genioglossus, mentalis, platysma and possibly even medial pterygoid, masseter and others too, as well as from the digastric. Thus a record made during chewing or opening with electrodes on the skin under the chin cannot be attributed solely to the digastric, without further evidence.

Another problem which arises when attempts are made to measure EMG is the extent to which it is possible to make quantitative comparisons between records from different individuals, or even between records from the same individual taken on different occasions. Because the voltages recorded depend upon the position and separation of electrodes, resistance of skin, thickness of subcutaneous fat, electrode and amplifier noise, stray pick-up from other sources as well as the activity of muscle fibres, they are subject to very large errors and therefore quantitative comparisons between measurements obtained under different conditions can be almost meaningless.

Some of these limitations can be overcome by using micro-electrodes in the form of needles (Basmajian and Stecko, 1962) or coiled wires (Bowman and Erickson, 1985) inserted into a muscle to pick up the all-or-none action potentials of individual muscle-fibres. Usually several muscle-fibres are innervated by one motor nerve - a motor unit- so that the all-or-none event which is recorded is the compound action potential from several fibres. Suitable electrodes can be made by cementing two thin insulated wires into a gauge 25 hypodermic or mandibular block needle and cutting off the wires flush with the bevel at the needle tip (Moyers, 1950; Milner-Brown et al., 1973; Hannam and Wood, 1981; Vitti and Basmajian, 1977). This type of electrode can be sterilized and inserted through the skin or mucous membrane into a muscle with little discomfort. A motor-unit is selected by moving the needle tip small distances within the muscle and once a suitable recording site has been found it is usually possible to hold the unit and record from it for some time, even while the muscle is shortening during a movement, although extensive movements could shift the needle tip and the unit would be lost. Recordings have to be made from a series of different units in turn to build up a picture of the properties of the whole population in the muscle.

The movement which the individual muscles are capable of producing can be deduced from examination of the cadaver, and these are described in textbooks of anatomy and several reviews (Carlsoo, 1952; Lindblom, 1960; Moller, 1966). However, it does not follow that each muscle is active in normal function during every movement to which it is capable of making a contribution.

Several studies have been done on the movements of the anterior region of the mandible during the masticatory cycle and all show that the pattern varies considerably between individuals, although any one individual usually has a characteristic pattern (Ahlgren, 1966; Hildebrand, 1931).

A much more detailed analysis of the masticatory movements would be possible in experimental animals and several different patterns of activity have been described on the basis of observations with cinefluorographic techniques (Crompton and Hotton, 1967; Hiiemae, 1966).

Gillings (1967) devised the photoelectric cell mandibulography for recording jaw movements in three dimensions.

Knap et al. (1970) studied jaw movement by placing a sensing device which consisted of six potentiometers arranged in precision linkage that provided electric signals for computer analysis, to the upper and lower jaws by cast clutches.

Araiche et al. (1966) and Goodson and Johansen (1975) used a mandibulograph which was capable of measuring the position of one point on the mandible in three dimensions, and was operated by electrical transducer measurement of the displacement of a rod connected through a flexible coupling to the lower incisor area. This six transducer mandibulograph produced quite useful data for characterization of the 3-D mandibular movement.

Karlsson (1977) and Hedegard (1979) recorded the mandibular movements by intraorally placed light emitting diodes.

The incidence of tooth contacts during mastication has been investigated very extensively and the most accurate information has been obtained by using micro radio transmitters incorporated into bridge pontics, which interfere minimally with normal movements (Glickman, 1968; Kavanagh, 1965).

Inhibition of the mandibular elevator muscles after maximum tooth contact in man either in the open-close, clench cycle (Griffin 1969, 1971) or during mastication (Ahlgren 1966, 1969, 1970) has also been reported.

Hannam et al. (1980) have measured jaw movements in three planes and recorded the associated electromyographic activity of six muscles during functional chewing movements. His technique consists of recording displacement by means of a transducer which senses the field strength of a small magnet cemented to the labial surface of the lower incisor teeth, the muscle activity was recorded by means of surface electrodes placed over the temporal and masseter muscles and using a computer to analyze all the data recorded.

Solomon and Waysenson (1979), have combined the advantage of the motion picture photography technique with the speed and accuracy of the electronic transducers without interfering with the patient's jaw during experiments by measuring the mandibular movements with a computer assisted radionuclide tracking technique.

Nevertheless, despite the sophistication of these recording techniques, the movements of the mandible represent the integrated result of the activity of the muscles of mastication and ligaments. The movement is regulated by an intricate neuromuscular system. From these observations, and studies on the neural mechanisms controlling jaw movements, it has been possible to obtain a better understanding of the mechanisms which regulate the function of the mandible in man.

4.1 Movements of the Temporomandibular Joint.

Sicher (1960) states that, "Mechanically speaking, the upper articulation is a sliding joint, that is, the disc and the condyle slide down and forward along the posterior slope and the flattening summit of the articular tubercle. The translatory movement may be executed symmetrically (forward thrust) or unilaterally (lateral swing). The two lower compartments together represent a hinge joint. The two discs are the socket of the "hinge" in which the mandible rotates." Sicher also adds that under normal conditions translation of the upper joint and rotation of the lower joints are always combined.

The Bennett movement is another movement occurring in the temporomandibular joint. It is generally regarded as a translatory movement occurring along the line of the hinge axis and not being of more than 2-3 mm, in extent. This movement only takes place on the working side of the articulation during lateral excursions. It would seem that the Bennett shift plays a part in locating the posterior area of the working side arch in a more lateral relationship to the maxillary arch thus facilitating a more uniformly cusp to cusp working occlusion. The movement takes place mainly in the upper joint compartment with both condyle and the meniscus shifting sideways.

Posselt (1952) placed emphasis on the limitations of joint movements by the capsule and its ligaments. He refers to the work of Langer (1865) and Meyer (1875) who state that the temporomandibular ligaments determine the course of the posterior opening movement in post-mortem operations, and also determine the path of this movement in the living individual.

Hjortsjo (1953) described opening movements of the jaws partially as rotation of the condyle around an axis in the vicinity of its own centre and partially as rotation of the temporomandibular meniscus around the articular eminence - these rotations have been referred to as capitulum and tuberculum rotations respectively. According to Hjortsjo (1953) the opening movement of the jaws starts with a rotation of the condylar heads and continues as a combination of capitulum and tuberculum rotation. At the termination of the opening movement tuberculum rotation predominates.

With regard to the actual relationships between the disc and articulating bones during movement, Rees (1954), has written an excellent paper. He discusses not only the anatomic shape of the meniscus but also relates this shape to function and postulates the movements between the condyle and the meniscus and also between the temporal bone.

He also states that "the temporo mandibular ligament is fairly taut in all positions of the joint and no doubt serves to keep condyle, disc and temporal bone firmly opposed. Limitation of forward movement, however, seems to result from the restraint offered by the posterior fibres of this ligament and of backward movement by the anterior fibres, while limitation of lateral movement results from the tension of the ipsilateral and of medial movements by the contralateral ligament".

The alternating movements of the mandible from side to side has been investigated by Sarnat (1951, 1980). Alternative methods of describing mandibular movements have also been proposed by Kraus et al. (1969).

It is important to note that both temporomandibular joints are necessarily involved in every jaw movement, forming together a bicondylar arrangement. Consult Hylander (1975) for further literature background and discussions.

4.1.1. Temporomandibular Articulation:

The temporomandibular joint is a compound joint. Its structure and function can be divided into two distinct systems:

a) One joint system is the tissues that surround the inferior synovial cavity. Since the disc is tightly bound to the condyle by the lateral and medial discal ligaments, the only physiologic movement that can occur between these surfaces is rotation of the disc on the articular surface of the condyle. Therefore the condyle-disc complex is the joint system responsible for rotational movement in the temporomandibular joint.

b) The second system is the condyle-disc complex, which functions against the surface of the mandibular fossa. Since the disc is not tightly attached to the articular fossa, free sliding movement can occur between these surfaces in the superior cavity. This movement occurs as a result of the mandible being positioned forward (translation). Translation therefore occurs in this superior joint cavity between the superior surface of the articular disc and condyle and acts as a nonorssified bone contributing to both joint systems, and hence the function of the disc justifies classifying the TMJ as a true compound joint.

The articular disc has been commonly referred to as a meniscus. However, it is not a meniscus at all. By definition, a meniscus is a wedge-shaped crescent of fibrocartilage attached on one side to the articular capsule and unattached on the other side, extending freely into the joint spaces.

A meniscus does not divide a joint cavity, isolating the synovial fluid, nor does it serve as a determinant of joint movement. Instead, it functions passively to facilitate movement between the bony parts. Typical menisci are found in the knee joint. In the TMJ the disc functions as a true articular surface in both joint systems and therefore more accurately is termed an articular disc.

The articular surfaces of the joint have no structural attachment or union, yet contact must be constantly maintained for joint stability. Stability of the joint is maintained by constant activity of the muscles that pull across the joint, primarily the elevators. Even in the resting state, these muscles are in a mild state of contraction called tonus. As muscle activity increases, the condyle is more greatly forced against the disc and the disc against the fossa, resulting in an increase in the interarticular pressure of these joint structures. It is possible that in the absence of interarticular pressure, the articular surfaces could separate and the joint might technically dislocate.

The width of the articular disc space varies with interarticular pressure. As the interarticular pressure increases, the condyle seats itself on the thinner intermediate zone of the disc. When the pressure is decreased and the disc space is widened, a thicker portion of the disc is rotated to fill the space. Since the anterior and posterior bands of the disc are wider than the intermediate zone, technically the disc could be rotated either anteriorly or posteriorly to accomplish this task. The direction of the disc rotation is determined by the structures attached to the anterior and posterior borders of the disc.

Attached to the posterior border of the articular disc are the retrodiscal tissues. The superior retrodiscal lamina is composed of elastic connective tissue. Therefore the effect of the superior retrodiscal lamina is to retract the disc posteriorly on the condyle. When the teeth are together and the condyle is in the closed joint position, the elastic traction on the disc is minimal. However, during mandibular opening, when the condyle is pulled forward down the articular eminence, the superior retrodiscal lamina becomes increasingly stretched, creating increased forces to retract the disc. In the full forward position the posterior retractive force on the disc created by the tension of the stretched superior retrodiscal lamina is at a maximum. The interarticular pressure and the morphology of the disc prevent the disc from being overretracted posteriorly. In other words, as the mandible moves into a full forward position and during its return the retraction force of the superior retrodiscal lamina holds the disc rotated as far posteriorly on the condyle as the width of the articular disc space will permit. This is an important principle in understanding the temporomandibular articulation. Likewise, it is important to remember that the superior retrodiscal lamina is the only structure capable of retracting the disc posteriorly on the condyle when the condyle is stationary.

Attached to the anterior border of the articular disc is the superior lateral pterygoid muscle. When this muscle is active, the disc is pulled anteriorly and medially. Therefore the superior lateral pterygoid is technically a protractor of the disc. This function, however, does not occur during most jaw movements. The superior lateral pterygoid is activated only in conjunction with activity of the elevator muscles during mandibular closure or a power stroke.

Like most muscles, the superior lateral pterygoid is constantly maintained in a mild state of contraction or tonus, which exerts slight anterior and medial force on the disc. In the resting closed joint position this anterior and medial force will normally exceed the posterior elastic retraction force provided by the nonstretched superior retrodiscal lamina. Therefore in the resting closed joint position, when the interarticular pressure is low and the disc space widened, the disc will occupy the most anterior rotary position on the condyle permitted by the width of the space. At rest with the mouth closed the condyle will normally be in contact with the intermediate and posterior zones of the disc.

This disc relationship is maintained during minor passive rotational and translatory mandibular movements. As soon as the condyle is moved forward enough to cause the retractive force of the superior retrodiscal lamina to be greater than the muscle tonus force of the superior lateral pterygoid, the disc is rotated posterior to the extent permitted by the width of the articular disc space. When the condyle is returned to the resting closed joint position, once again the tonus of the superior lateral pterygoid becomes the predominant force and the disc is repositioned forward as far as the disc space will permit.

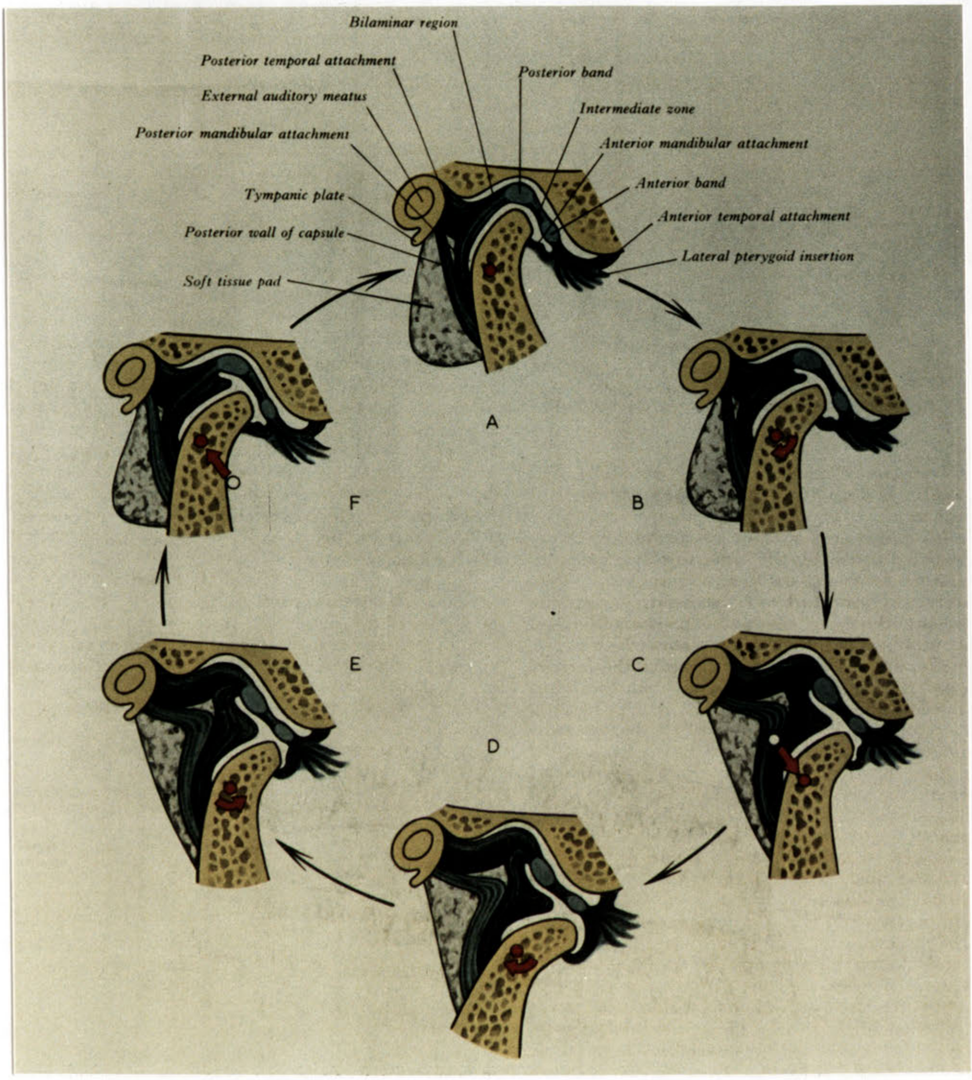


Figure 32 Normal functional movement of the condyle and disc during the full range of opening (A-D) and closing (D-A) (Rees, 1954).

During chewing, when resistance is met during mandibular closure, such as when biting on hard food, the interarticular pressure on the biting side is decreased. This occurs because the force of closure is not applied to the joint but is instead applied to the food. With the condyle forward and disc space increased, the tension of the superior retrodiscal lamina will tend to retract the disc from a functional position. This will cause separation of the articular surfaces, resulting in a dislocation. To avoid this, the superior lateral pterygoid becomes active during the power stroke, rotating the disc forward on the condyle so the thicker posterior border of the disc maintains articular contact. Therefore joint stability is maintained during the power stroke of chewing. As the teeth pass through the food and approach intercuspation, the interarticular pressure is increased. As the pressure is increased, the disc space is decreased and the disc is mechanically rotated posteriorly so the thinner intermediate zone fills the space. When the force of closure is discontinued, the resting closed joint position is once again assumed.

4.2 Mandibular Movements:

The masticatory muscles combine in various patterns to execute the different movements of the mandible. It is especially important to realize that one muscle may act synergistically with different muscles at different times. In no instance does a muscle act as an independent unit; instead muscles act in groups and almost always in suprisingly large groups. In addition, a single muscle may have portions that can function differently. This is particularly true of the temporalis and lateral pterygoid muscles.

The most important muscles that affect movements of the mandible can be divided into three exclusive groups:

i) Elavators, or the temporalis, masseter and medial pterygoid muscles;

ii) Depressors, or the digastric, mylohyoid and geniohyoid muscles; and

iii) Protractors, the lateral pterygoid muscles.

The retractors of the mandible do not constitute an independent group. They are represented by the middle and posterior fibres of the temporalis muscle, the deep portion of the masseter, and the mylohyoid, geniohyoid and digastric muscles. Thus, both elevators and depressors may function as retractors of the mandible.

4.2.1. Mandibular Rest Position:

It is commonly assumed that the mandible is maintained in a fairly constant position in relation to the maxilla when a person is relaxed, and that the maintenance of this position is the function of some postural control mechanism which produces a low level of activity in the elevator muscles to oppose the effects of gravity (Posselt, 1963).

If the elevator muscles were engaged in actively maintaining the position of the mandible against gravity in the rest position, as has usually been assumed, it should be possible to record EMG activity from them. The evidence on this however, is not conclusive one way or the other. There have been reports of low levels of activity in the masseter, temporalis, lateral pterygoid and suprahyoid muscles at rest (Carlsoo, 1952, 1958; Kawamura et al.,1967; Lehr et al., 1971; Lund et al., 1970) and, furthermore that there are changes in the activity of these muscles when the body is tilted, suggesting that the resting activity may be adjusted according to gravitational effects (Lund et al., 1970). However, no actual measurements were made to show that the mandibular position remained constant in these experiments.

A total absence of muscle activity at rest would not automatically exclude the existence of a control system regulating the resting posture of the mandible, however. It might be that the equilibrium position due to just the passive forces of the muscles, ligaments and joint capsules coincides with the position "sought" by the control system, in which case the muscles would only be called into action to correct any transient imbalance or to oppose some external force which tended to disturb the passive equilibrium. Such a mechanism is thought to operate in many of the limb and trunk muscles in maintaining an erect posture during standing.

4.2.2 Movements of Initial and Maximal Opening:

During simple opening and closing mandibular movements, the mandible rotates around a frontal axis that passes approximately through the centres of the two condyles while the axis itself progresses in space (Ulrich 1886, Gibbs 1969).

Moyers (1950) in his EMG analysis of temporomandibular movements postulated that the opening movement is caused by a combined action of the lateral pterygoid and digastric muscle. Today it is well known that the action of geniohyoid and mylohyoid cannot be excluded from this movement. If this movement occurs without resistance, the depressors act without any great force. The protracting force of the inferior head of the lateral pterygoid muscles acting upon the condyles and discs, the depressing and retracting force of the geniohyoid and digastric muscles acting upon the chin, and the action of the mylohyoid muscle upon the body of the mandible combine to produce rotatory and translatory jaw movement.

The movement of the mandible from centric occlusion to the resting position has been termed by Lindblom (1960) "The small translatory movement of the jaws". Attempts to measure the condylar translation directly on radiographs have been made, by Higley and Logan (1941), Beyron (1942) and Lindblom (1960). Posselt (1952) had stated that the change of position of the mandible from centric occlusion to the rest position was in the nature of a bodily movement.

The large opening movement of the jaws is agreed by most authors to be achieved by a combination of rotatory and translatory movements of the mandibular condyles (Sarnat, 1964, 1980).

The movement which starts from centric occlusion commences as a vertical drop (translation, slide). From this position the movement continues as a forward translation to near the anterior part of the articular eminence. During this part of the opening movement there appears to be only small transversal rotatory movements. However, the terminal part of opening is achieved mainly by a rotatory (hinge) movement of the condyles (Lundberg 1963).

According to Sicher (1960) and Hylander (1975) in the opening movement of the jaw the condyles rotate against the discs around a transverse axis as they slide downward and forward along the posterior slope and summit of the articular eminence. The movement is effected by an initial activity of the lateral pterygoids, which first fix the condyles firmly against the posterior slope of the eminence. Immediately this is followed by contraction of the digastric muscles, and the sustained activity of both muscle pairs, acting as a force-couple, turns the mandible around a roving horizontal axis passing through the rami of the mandible. This motion affects all the other muscles anchored to the mandible, but it has also a receding influence on muscles and more peripheral to the central action. Thus the elevators of the jaw must lengthen to act as mild balancers and so ensure smoothness of performance.

According to some authors the oblique portion of the temporomandibular ligament also influences the normal opening movement of the mandible. During the initial phase of opening, the condyle can rotate around a fixed point until the temporomandibular ligament becomes tight as its point of insertion on the neck of the condyle is rotated posteriorly. When the ligament is taut, the neck of the condyle cannot rotate further. If the mouth were to be opened wider, the condyle would need to move downward and forward across the articular eminence.

This unique feature of the temporomandibular ligament which limits rotational opening, is found only in humans.

In the erect postural position and with a vertically placed vertebral column, continued rotational opening movement will cause the mandible to infringe on the vital submandibular and retromandibular structures of the neck. The outer oblique portion of the temporomandibular ligament possibly functions to resist this infringement.

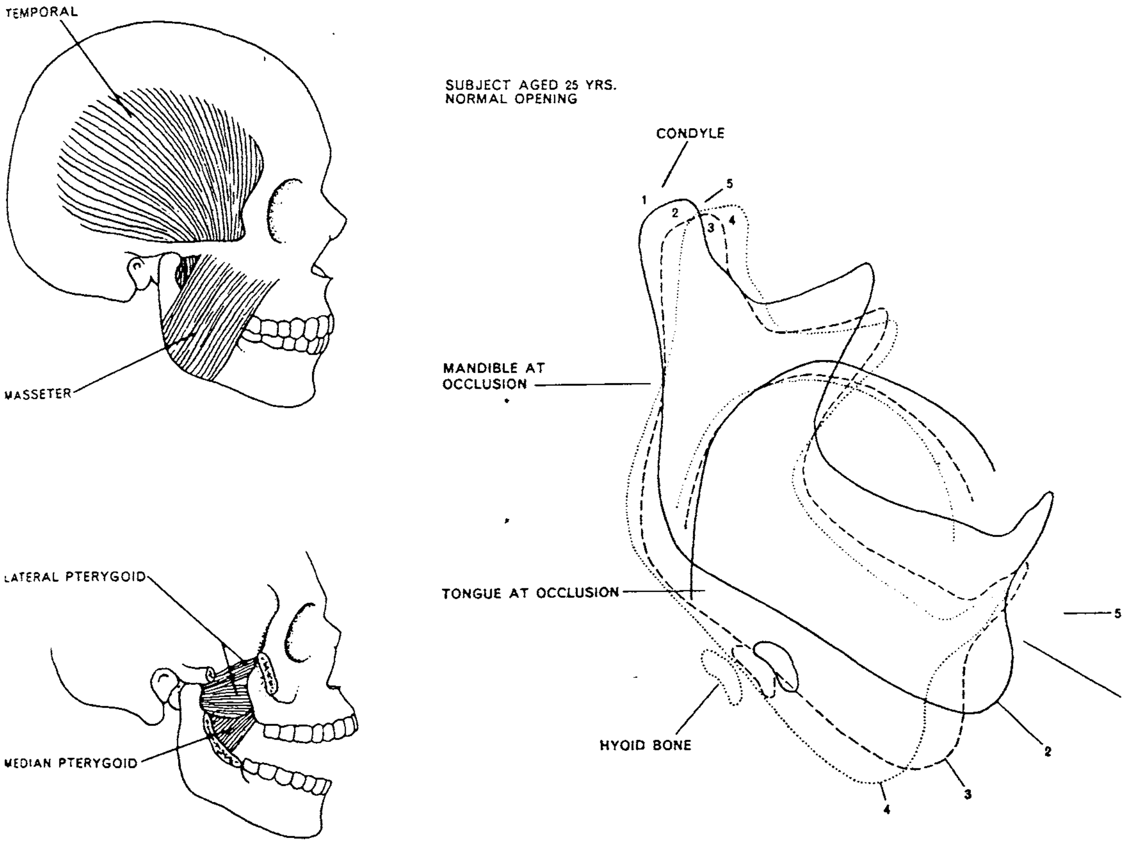


Figure 33 Normal Opening Movement of the Mandible Composite of functional cephalometric radiographs.

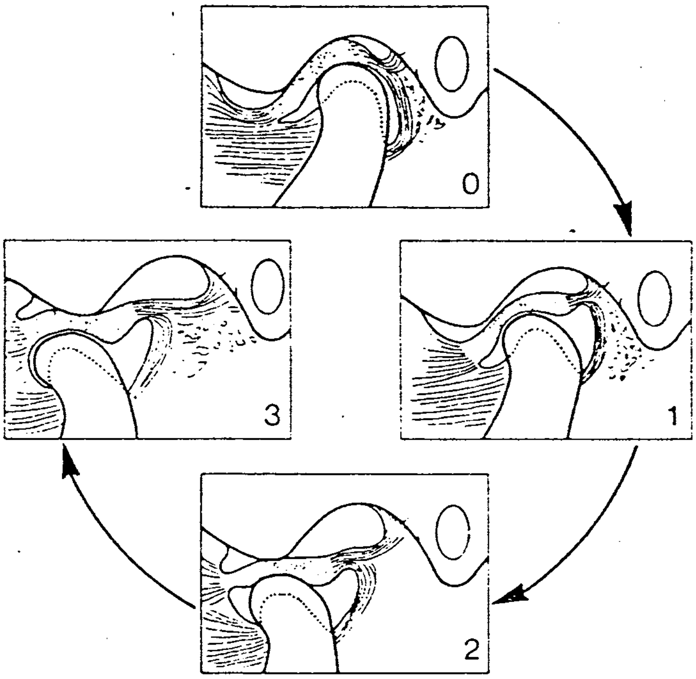


Figure 34 The normal TMJ hard and soft tissue articulating relationship during opening movement.The four drawings represent the relationships of the disc to bony structures a closed position and various stages of opening, as seen arthrographically.(from Wilkes, C. H.,1978).

4.2.3 Lateral Movements of the Jaws

Lateral movements of the jaws are assymmetrical. The side to which the mandible moves is called the working side (Ramfjord and Ash, 1971). The opposite side is called the balancing side or the non-functional side. In general, the axes of right and left lateral movements are located posterior to the working side condyles (Fisher, 1952). Because of the position of the axis for lateral movements of the jaws the working side condyle will shift laterally in the direction of movement and probably slightly downwards (Bennett, 1908). In the average individual the Bennett shift is approximately 1.5 mm in the lateral direction corresponding to edge - to - edge position of the cuspids on the working side (Posselt, 1962). The balancing side condyle during lateral excursion moves downwards, forwards and medially. The angle traced by the balancing side condyle in relation to the sagittal plane is called Bennett angle. The downward movement of the balancing side condyle is greater than the downward movement of the working side condyle and because of this the distance between the dental arches will be greater on the balancing side than the working side (Christensen, 1905).

Lateral movements of the mandible may be inspected as regards habitual contact gliding movements from centric occlusion to either side or alternatively as non-contact lateral movements.

Some observers (Beyron, 1969; Krogh-Poulsen and Olsson, 1969) are of the opinion that several contacts are desirable between the teeth on the functional side during lateral excursion whilst other observers are of the opinion that there should only be contact between the upper and lower canines on the functional side (D'Amico, 1958; Stallard and Stuart, 1963; Stuart, 1964). The latter concept of contact lateral gliding movements of the jaws have been called canine protection and asserts that contacts between the posterior teeth should only occur in centric occlusion and that during lateral excursion all teeth except the canines on the functional side should immediately be taken out of occlusion by contact between the gliding inclines of the canines.

According to Sicher and Du Brul (1975), "in the lateral movement one condyle and disc slide downward, medially, and forward along the articular eminence while the other rotates laterally around a vertical axis". The lateral pterygoid muscle, inserted on the inwardly thrust medial pole of the condyle, pulls inward and forward in the horizontal plane. The horizontal fibres of the temporalis muscle, inserted at the posterior tip of the coronoid process, pull outward and backward.

These muscles, operating as a force-couple, contribute to the torque of the rotating condyle necessary to effect chewing on this side. But perhaps equally important, they let out slack with a strong lengthening tension to provide maximum stability of the condyle.

If the chin is moved to the left, the holding posterior part of the temporalis muscle forces the left condyle towards its marginal or extreme position under the restricting influence of the temporomandibular ligament aiding in the Bennett Shift. At the same time the right pterygoid muscle pulls the right condyle and disc forward, downward, and medially as the whole jaw is swung, as well as shifted bodily, to the left.

In the closing stroke the force-couple changes in direction and components. The posterior fibres of the right temporalis move the mandible back toward the right while the posterior fibres of the left temporalis hold. At the same time both pterygoids lengthen to balance, the right one having by far the greater excursion. The right digastric contracts to aid this movement and the cranial and hyoid bone holders are also active participants.

4.2.4. Closing Movement.

The closing movement is executed by the elevators of the mandible. If the mouth is opened to its maximal extent, the timing of the activation and relaxation of the different parts of these muscles is important for proper closure.

Each disc and condyle glides anteriorly to the summit of the articular eminence in maximal opening. This physiological dislocation of the temporomandibular joint must be eliminated before closure can proceed. Therefore, contraction of the elevating component is delayed until the condyles and discs have regained their position at or behind the height of the articular eminence. Thus, the first phase of the closing movement from this position involves the combined action of the retracting portions of the masseter and temporalis muscles.

When the first phase of the movement has been executed (in which the mandible glides sharply backward without rotating upward to an appreciable extent), the closing movement is completed by the elevators, which return the jaw to the rest or occlusal position.

4.2.5. Protrusion and Retrusion Movements.

Moyers (1950), Woelfel et al. (1960) and Moller (1966) showed that forward thrust of the mandible is the result of contraction of the inferior head of the lateral pterygoid muscles and holding action of the masseter and medial pterygoid muscle. The temporalis muscle is not active during this movement, and the depressors are only minor active.

The lateral pterygoid pull the mandibular condyles and discs forward and downward along the articular eminences, while the elevators and depressors apparently maintain the position of the mandible relative to the maxilla, thereby preventing the lower jaw from dropping.

In the retracting movement, the middle and posterior fibres of the temporalis muscle combine forces with the depressors, while the remaining elevators exhibit varying amounts of activity. The depressing component of the suprahyoid muscles is apparently neutralized by the activity of the temporalis muscle.

4.2.6. Masticatory Movements

The masticatory movements of the lower jaw frequently occur under considerable force and often lead to contact between the upper and lower teeth (Sicher, 1954; Anderson, 1957; Ahlgren, 1966). When food is initially introduced into the mouth, few contacts occur. As the bolus is broken down, the frequency of tooth contacts increases. In the final stages of mastication, just prior to swallowing, contacts occur during every stroke (Adams and Zander, 1964).

Mastication is defined as the act of chewing foods. It represents the initial stage of digestion, when the food is broken down into small particle sizes for ease of swallowing.

Mastication is made up of rhythmic and well controlled separation and closure of the maxillary and mandibular teeth. Each opening and closing of the mandible represents a chewing stroke. The complete chewing stroke has a movement pattern described as tear sharped. It can be divided into opening phase and closing phase. The closing movement has been further subdivided into the crushing phase and the grinding phase. During mastication similar chewing strokes are repeated over and over as the food is broken down.

Although this general movement pattern exists, the actual masticatory movements vary in detail both within and between individuals. These movements are in part dependent on the shape and proportions of the jaws and teeth and the type of food masticated. In general, individuals with malocclusions exhibit more complex and irregular chewing patterns (Ahlgren, 1966).

When the mandible is traced in the frontal plane during a single chewing stroke, the following sequence occurs:

In the "opening" phase it drops downwards from the intercuspal position to a point where the incisal edges of the teeth are about 16 to 18mm apart (Hildebrand, 1931). It then moves laterally 5 to 6mm from the midline as the closing movement begins. The first phase of closure traps the food between the teeth and is called the "crushing" phase (Fig. 35).

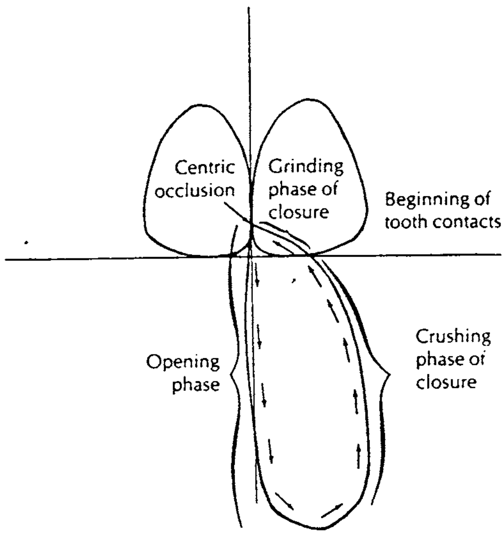


Figure 35: Frontal view of the chewing stroke.

As the teeth approach each other, the lateral displacement is lessened so that when the teeth are only 3mm apart the jaw occupies a position only 3 to 4mm lateral to the starting position of the chewing stroke (Hilderbrand, 1937). At this point the teeth are so positioned that the buccal cusps of the mandibular teeth are almost directly under the buccal cusps of the maxillary teeth on the side to which the mandible has been shifted.

As the mandible continues to close, the bolus of food is trapped between the teeth. This begins the grinding phase of the closure stroke.

During the grinding phase the mandible is guided by the occlusal surfaces of the teeth back to the intercuspal position, which causes the cuspal inclines of the teeth to pass across each other, permitting shearing and grinding of the bolus of food.

If movement of the mandibular incisor is followed in the sagittal plane during a typical chewing stroke, it will be seen that during the opening phase the mandible moves slightly anteriorly. During the closing phase it follows a slightly posterior pathway, ending in an anterior movement back to the maximum intercuspal position. More detailed analysis of the movements of Mastication is given by Lundeen and Gibbs (1982).

The muscle activity patterns during mastication are fairly well known(Moller, 1966). The opening stroke of mastication is initiated first by activity of the depressor group, with the mylohyoid muscles contracting somewhat earlier than the digastric muscles. The mandible is depressed as a result of this activity. Then, the inferior heads of the lateral pterygoid muscles contract, with activity on the chewing or ipsilateral side ordinarily preceding activity on the contralateral side. As a result, the mandible is rotated open and translated forward and slightly away from the chewing side. The mandible moves toward the chewing side when levels of activity in the lateral pterygoid muscle on the contralateral side exceed those of the ipsilateral muscle.

The closing stroke is initiated by contraction of the medial pterygoid muscles. At this time, the chewing side of the mandible is still being moved laterally by the inferior head of the contralateral pterygoid muscle. Following initiation of medial pterygoid muscle activity, the ipsilateral temporalis muscle becomes active, followed by activity in the remaining elevators. The combined effect of the elevators is closing of the jaws. The lateral pterygoid muscles are only slightly, if at all, active during the end of the closing stroke.

The closing stroke grades into the power stroke. The size of the food object determines when the power strokes occurs. When the teeth are in forciable contact with food and with one another, the electromyographic activity reaches a maximum (Moller, 1966). This has been interpreted as the period of maximum occlusal force. During the power stroke, the lateral pterygoid muscles become active, with the ipsilateral side exhibiting more activity than the contralateral side. The activity of the lateral pterygoid muscle during the power stroke is confined to its superior head (Lipke et al., 1977; McNamara Jr., 1973).

If the chopping jaw movement is the same as puncture-crushing, electromyographic data suggest that there are larger forces being generated during tooth-tooth contact of the power stroke than during puncture-crushing (chopping) (Ahlgren, 1966).

This data also suggests that the activity in the ipsilateral temporalis muscle precedes activity in the contralateral temporalis and masseter muscles during tooth-tooth contact, while these four muscles contract simultaneously during chopping.

Puncture-crushing or chopping is primarily a series of up and down vertical strokes. The elevators contract simultaneously during chopping. This results in the common observation that mandibular movement during this type of power stroke is primarily directed vertically.

The amount of force placed on the teeth during mastication varies greatly from individual to individual and it is related to physiology and neuroanatomy of the muscles.

All functional movements explained are highly co-ordinated complex neuromuscular events. Sensory input from the structures of the masticatory system is received and integrated with existing reflex actions and learned muscle activity to achieve a desired functional activity.

CHAPTER V

FUNCTIONAL NEUROANATOMY

The function of the masticatory system is complex. Discriminatory contraction of the various head and neck muscles is necessary to move the mandible precisely and allow effective functioning. A highly refined neurologic control system regulates and co-ordinates activities of the entire masticatory system. It consists primarily of nerves and muscles; hence the term neuromuscular system.

5.1 Muscle Structure

The skeletal muscles and the joints are designed to fulfill the body's need for mobility and stability. A human skeleton without muscles will collapse when placed in the erect standing position. The two types of materials found in skeletal muscle are connective tissue and muscle tissue. The properties of these tissues and the way in which they are interrelated give muscles their unique characteristics. Muscle tissue like other biologic tissue is viscoelastic. It also possesses the properties of contractility and iritalibity. Contractility is the muscle's ability to develop tension against resistance. Irritability refers to a muscle's ability to respond to chemical, electrical or mechanical stimuli.

The connective tissue in muscle is dense fibrous tissue in the form of sheaths and tendons. The sheaths provide the supporting framework for the muscle tissue, while the tendons serve to attach the muscles to the bones and to transmit forces generated by the muscle to the bones.

A skeletal muscle is composed of many thousands of muscle fibres. The arrangement, number, size and type of these fibres may vary from muscle to muscle (Johnson and Pogar, 1973), but each fibre is a single muscle cell that is enclosed in a cell membrane called the sarcolemma. The muscle fibre is composed of cytoplasm, which is called sarcoplasm. The cell nuclei, mitochondria, myoglobin, glycogen, and myofibrils are contained within the sarcoplasm.

A single muscle fibre may contain many myofibrils, and each myofibril contains tiny filaments called myofilaments. These filaments are composed of the proteins actin and myosin. When the myofibril is viewed through a microscope, the alternation of thick myosin and thin actin filaments forms a distinctive striped pattern. Therefore skeletal muscle is often called striated muscle.

The sarcomere is considered to be the contractile element of a muscle. When a muscle is stimulated, the chemical changes that occur within the sarcomere cause the actin filaments to slide. As the actin slides past the myosin, a chemical reaction binds actin to myosin, creating the complex actomyosin. This bonding of filaments is called a crossbridge, and is possible only at specific bonding sites of the filaments. The contraction of a muscle is the result of many contractile units undergoing sliding of filaments and crossbridge formations.

Three principal types of fibres are found in varying proportions in skeletal muscles. These fibre types may be distinguished from one another histochemically, metabolically, morphologically, and mechanically (Rosse and Clawson, 1980)The terminology that has been used also differ with different authors. The mostcommonly accepted terminology to designate the fibres is fast twitch glycolytic (FG), fast twitch oxidative glycolytic (FOG), and slow twitch oxidative (SO) (Taylor, 1980).

Skeletal muscles are composed of all three types of fibres, but wide variations exist among individuals in the fibre composition of various muscles. The relative proportions of different fibre types in a muscle determine in part the function of the muscle. Muscles that have a relatively high proportion of SO fibres in relation to FG fibres, are able to carry on sustained activity because the SO fibres do not fatigue rapidly. Muscles that have a high proportion of the FG fibres, can respond more rapidly to a stimulus than SO fibres.

The histochemical data relevant to jaw muscles are limited (Taylor, 1976). All accounts agree that the jaw-closing muscles are heterogeneous. With respect to myosin ATPase (MATPase),Ringqvist (1973) recognized three levels of staining in human biopsies from masseter and temporalis. Weak and strong staining fibres were identified with types I and II of biceps brachii, but moderately staining ones were believed to be a special group which was called intermediate. Later they were modified by Ariano et al. (1973) and Taylor et al. (1973) to type B (I), type A (II) and C (intermediate). The types A and C originally were defined by Stein and Podykula (1962).

In their scheme, A is equivalent to fast glycolytic (FG), B to slow oxidative (SO) and C to fast oxidative glycolytic (FOG) of Ariano et al. (1973).

Ringqvist (1973a, 1973b,1974) published a data related to the types of fibres and mean fibre cross sectional areas of three jaw closing muscles in humans. Taylor et al. (1973) compared the results with adult cats. They have found that the most striking difference between human and cat fibre type distribution was in the proportions of type B. (Table.II)

The quantitative significance of the proportions of the various types could be seen in estimates of their contribution to the cross-sectional area of the muscle, since this could indicate their approximate relative force generating capabilities.

TABLE II

Proportions of fibre types in human and cat-closing muscles (Taylor et al. 1976)

	A%	B%	C%	A+C%	A/C%
HUMAN:					
Masseter	58	26.4	15.6	73	3.7
Temporalis	23	42	25	48	0.92
CAT:					
Masseter	82	10	8	90	10.3
Pterygoid	59	29	20	71	2.6
Temporalis	72	2	26	98	2.8

TABLE III

Estimated mean fibre cross-sectional areas and percentage areas for the fibre types in human jaw closing muscles (Ringqvist, 1973; Taylor et al. 1973)

	A	B	C	A+C	A/C
Mean Areas μm^2					
MASSETER	201	830	419	--	--
TEMPORALIS	377	1359	499	--	--
Estimated % Areas :					
Masseter	29.0	54.6	16.4	45.4	1.8
Temporalis	32.2	65.4	2.4	34.6	13.4

While these fibres histochemically can be distinguished the morphological differences is also very important. It is well known that the tension developed by a muscle varies with its length. The contractile unit of a skeletal muscle fibre is a sarcomere, and above and below the optimum sarcomere length, the tension developed during a contraction declines.

The sarcolemma of an individual muscle fibre is surrounded by connective tissue called endomysium, and bundles of muscle fibres are covered by connective tissue called endomysium, and groups of muscle fibres are covered by connective tissue called perimysium (Fig. 36).

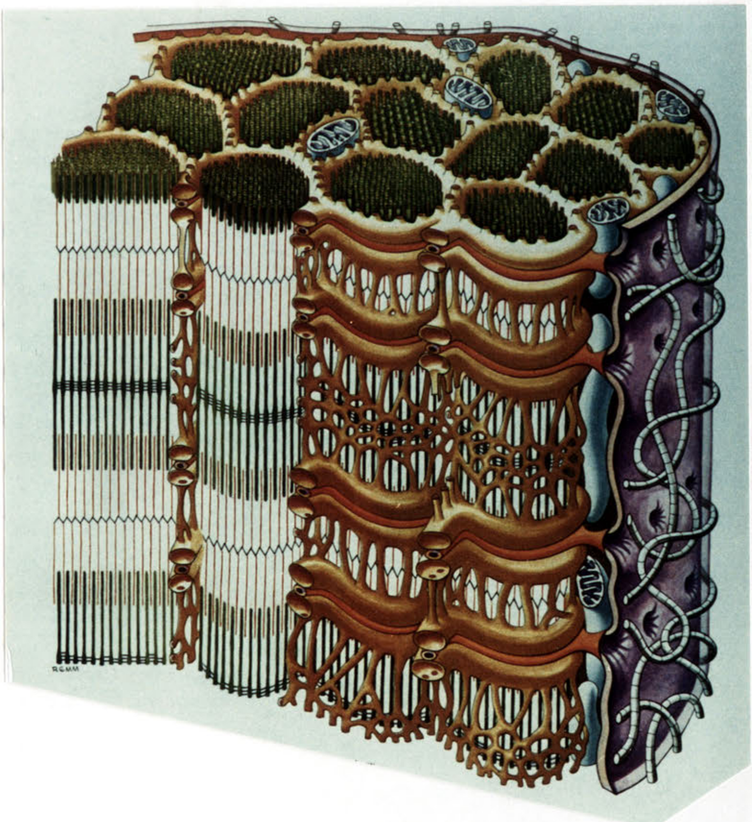


Figure 36 Three dimensional reconstruction of skeletal muscle (Gray's Anatomy, 1982).

The groups or bundles of muscle fibres are called fasciculi. The endomysium and perimysium are continuous with the outer connective tissue sheath called the epimysium which envelopes the entire muscle. Continuations of the outer sheath form the tendons that attach each end of the muscle to the bony components. The tendons are attached to the bones by Sharpey's fibres. The connective tissues that surround the muscle fibres are non parallel with the muscle fibres. These tissues, plus the sarcolemma, mitochondria, and other structures (nerves and blood vessels), form the parallel elastic component of a muscle. This means that when a muscle shortens, these tissues will also shorten or act together with the muscle fibres. The tendon of the muscle, on the other hand, is in series with the contractile elements and when the muscle actively shortens, the tendon will be stretched.

The length, arrangement, and number of muscle fibres vary from muscle to muscle throughout the body. These structural variations affect not only the shape and size of the muscles but also the function of the various muscles (Yemm, 1976, 1977).

Muscle fibres are capable of shortening to approximately one-half of their total length.

Arrangement of fasciculi may be parallel, at an angle, or spiral around the long axis. Muscles that have a parallel fibre arrangement are designated as strap or fusiform muscles. Generally muscles with a parallel fibre arrangement will produce a greater range of motion of a bony lever at a joint than muscles of similar size with different fibre arrangements.

Muscles that have an oblique fibre arrangement are called unipennate, bipennate, or multipennate muscles because the fibre arrangement resembles that found in a feather. The fibres that make up the fasciculi in pennate muscles are usually shorter and more numerous than the fibres in many of the strap muscles. Pennate muscles are generally unable to produce a large range of motion however they are capable of developing a large amount of force.

5.2 The Muscle Function:

In considering the contraction of a muscle as a whole it is important to understand how its individual elements operate. The basic component of the neuromuscular system is the motor unit, which consists of a number of muscle fibres that are innervated by one motor neuron.

The size of motor units varies between muscles, smaller units occurring where precise control of muscular action is required.

Physiological investigations have shown that the muscle fibres innervated by one motor neuron are often widely spread within a muscle and do not necessarily correspond to its myonemes or fascicular divisions. Thus, even when only a few motor units are active, the force is generated diffusely. It has also been shown that each axon terminates within a round or oval territory, with considerable overlap between adjacent zones of innervation.

When the neuron is activated, volleys of nerve impulses pass along its membrane, and, reaching the terminal, cause the liberation of a chemical transmitter, acetylcholine, close to the adjacent muscle fibre membrane. The transmitter is then rapidly broken down by anzymic action so that it operates only briefly. Electrical recordings show that the transmitter causes changes in the permeability of, and hence the electrical potential across, the sarcolemma. In resting fibres the so-called resting potential has a steady value of about 80 mV, the interior of the cell being negative with respect to the exterior. When stimulated by low concentration of acetylcholine, a small transient localized depolarization of the membrane, the end plate potential occurs. This does not itself cause contraction, but with increased concentrations of transmitter it builds up until a critical level is reached, at which point the neighbouring membrane abruptly undergoes a massive rapid depolarization, continuing to reversal of the membrane potential so that the interior becomes 20 mV positive, relative to the exterior. This is known as the action potential; the swing rapidly reverts to the resting potential but it initiates a wave of depolarization, potential reversal and return to resting levels, which sweeps at a speed of 5 metres per second over the whole surface of the muscle fibre, and this causes its momentary contraction. The mechanical response is a single twitch lasting 25-75 msec. Repeated action potentials can be elicited from a single muscle fibre if the activity of its motor axon is maintained, but there is a maximum limit to this since, after a single muscle action potential has occurred, there is a refractory period of about 10 msec until another can be elicited.

Since all action potentials are identical, as are all isolated twitches of a given muscle fibre, they are said to show an "all or none" response, if they contract at all, they contract maximally within the limits imposed by their initial length and conditions of loading.

Whole muscles, however, exhibit considerable gradation in their contraction and this is achieved by differential activity of the motor units.

Individual units can vary greatly in their twitch frequency, and the number of units that are active also fluctuates. In small contractions only a few units are operative, but with increasing contraction more are recruited until many or all are active. The sum of these activities results in a steady contraction of the whole muscle even though the individual units are twitching repetitively, but in an a synchronous manner.

The number of muscle fibres innerated by one motor neuron varies greatly according to the function of the motor unit. The fewer the muscle fibres per motor neuron, the more precise is the movement. A single motor neuron may innervate only two or three muscle fibres. Conversely one motor neuron may innervate hundreds of muscle fibres as in any large muscle. There is a similar variation in the number of muscle fibres per motor neuron within the muscles of mastication.

The lateral pterygoid muscle has a relatively low muscle fibre/motor neuron ratio and therefore is capable of fine adjustments in length needed to adapt to horizontal changes in the mandibular position. By contrast, the masseter has a greater number of motor fibres per motor neuron, which corresponds to its more gross functions of providing the force necessary during mastication.

The motor unit can carry out only one action : contraction or shortening. The entire muscle, however, has three potential functions.

i) When a large number of motor units in the muscle are stimulated, contraction or an overall shortening of the muscle occurs. This type of shortening under a constant load is called isotonic contraction. Isotonic contraction occurs in the masseter when the mandible is elevated, forcing the teeth through a bolus of food.

ii) When a proper number of motor units contract opposing a given force, the resultant function of the muscle is to hold or stabilize the jaw. This contraction without shortening is called isometric contraction, and it occurs in the masseter when an object is held between the teeth.

iii) A muscle also can function through controlled relaxation. When stimulation of the motor unit is discontinued, the fibres of the motor unit relax and return to their normal length. By control of this decrease in motor unit stimulation, a precise muscle lengthening can occur that allows smooth and deliberate movement. This type of controlled relaxation is observed in the masseter when the mouth opens to accept a new bolus of food during mastication.

5.3 Neurologic Structures

Each skeletal muscle has both sensory and motor innervation. The sensory or afferent neurons carry information from the muscle to the central nervous system at both the spinal cord and the higher centre levels. The type of information carried by the afferent nerve fibres most often depends on the sensory nerve endings. Some nerve endings relay sensations of discomfort and pain, as when the muscle is fatigued or damaged. Others provide information regarding the state of contraction or relaxation of the muscle. Still others provide information regarding joint and bone positions (proprioception).

Once the sensory information has been received and processed by the central nervous system, regulatory information is returned to the muscles by way of the motor or efferent nerve fibres. The efferent neurons indicate the impulses for the appropriate function of the specific muscles that will bring about the desired motor response.

Sensory receptors are neurologic structures or organs located in the tissues that provide information to the central nervous system regarding the status of these tissues. As in other areas of the body, various types of sensory receptors are located throughout the tissues that make up the masticatory system. There are specialized sensory receptors that provide specific information to the afferent neurons and thus back to the central nervous system. Constant input received from them allows the brain to co-ordinate action of individual muscles or muscle groups so that smooth finely adjusted movements can occur.

Like other systems, the masticatory system utilizes four major types of sensory receptors to monitor the status of its structures:

i) the muscle spindles, which are specialized receptors found in the muscle tissue (Granit, 1955);

ii) the Golgi tendon organs, located in the tendons (Granit, 1955);.

iii) the pacinian corpuscles, located in tendons, joints, periosteum, fascia, and subcutaneous tissues; and

iv) the neciceptors, found generally throughout all the tissues of the masticatory system.

5.3.1. Muscle Spindles:

Skeletal muscles consist of two types of muscle fibre, the first is the extrafusal fibres, which are contractible and make up the bulk of the muscle; the other is the intrafusal fibres, which are only minutely contractile.

A bundle of intrafusal muscle fibres bound by a connective tissue sheath is called a muscle spindle. The muscle spindles are interspersed throughout the skeletal muscles and aligned parallel to the extrafusal fibres. Within each muscle spindle the nuclei of the intrafusal fibres are arranged in two distinct fashions: chainlike (nuclear chain type) or clumped (nuclear bag type), (Fig.37).

There are two types of afferent nerves that supply the intrafusal fibres. They are classified according to their diameters. The larger fibres conduct impulses at a higher speed and have lower thresholds. Those that end in the central region of the intrafusal fibres are the larger group (Ia) and are said to be the primary endings (annulospiral endings). Those that end in the poles of the spindle are the smaller group (II) and are the secondary endings (flower spray endings) (Karlsson et. al., 1971, 1974,).

Since the intrafusal fibres of the muscle spindles are aligned parallel to the extrafusal fibres of the muscles, as the muscle is stretched so also are the intrafusal fibres. This stretch is monitored at the nuclear chain and nuclear bag regions. The annulospiral and flower spray endings are activated by the stretch, and the afferent neurons carry this information to the central nervous system. The afferent neurons originating in the muscle spindles of the muscles of mastication have their cell bodies in the trigeminal mesencephalic nucleus (Scalzi and Prince, 1972).

The intrafusal fibres receive efferent innervation by way of fusimotor nerve fibres (Barker et al. 1972). These fibres are given the classification of gamma fibres or gamma efferents to distinguish them from the alpha nerve fibres, which supply the extrafusal fibres.

When the intrafusal fibres contract, the nuclear chain and nuclear bag areas are stretched, which is registered as if the entire muscle were stretched, and afferent activity is initited. Thus there are two manners in which the afferent fibres of the muscle spindles can be stimulated:

i) generalized stretching or elongation of entire muscle (extrafusal fibres), and

ii) contraction of the intrafusal fibres by way of gamma efferents.

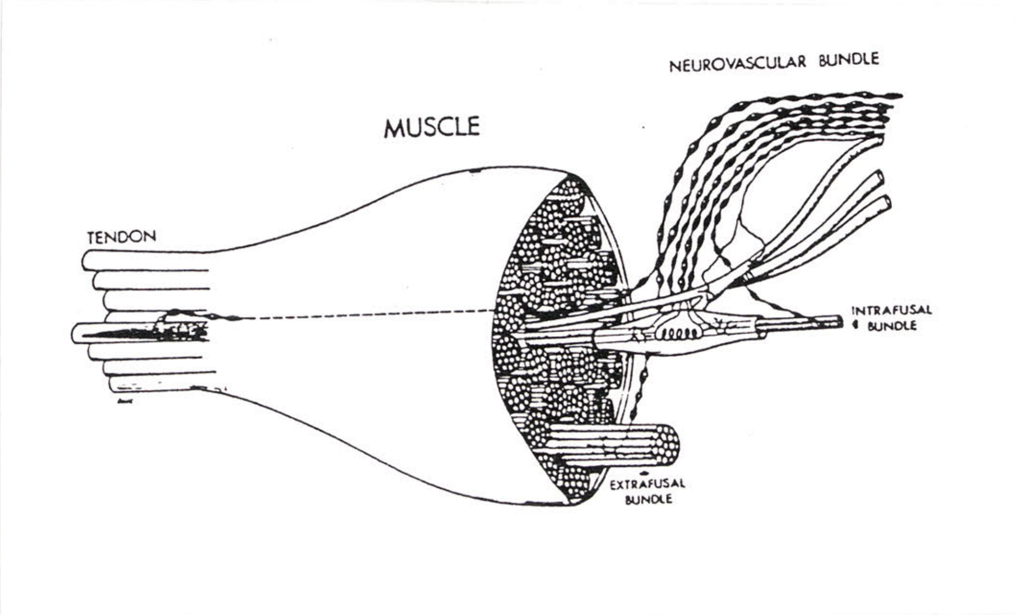


Figure 37 Diagram of the relationship of a muscle spindle and a classic tendon organ to the extrafusal muscle bundles and neurovascular bundle of a muscle.

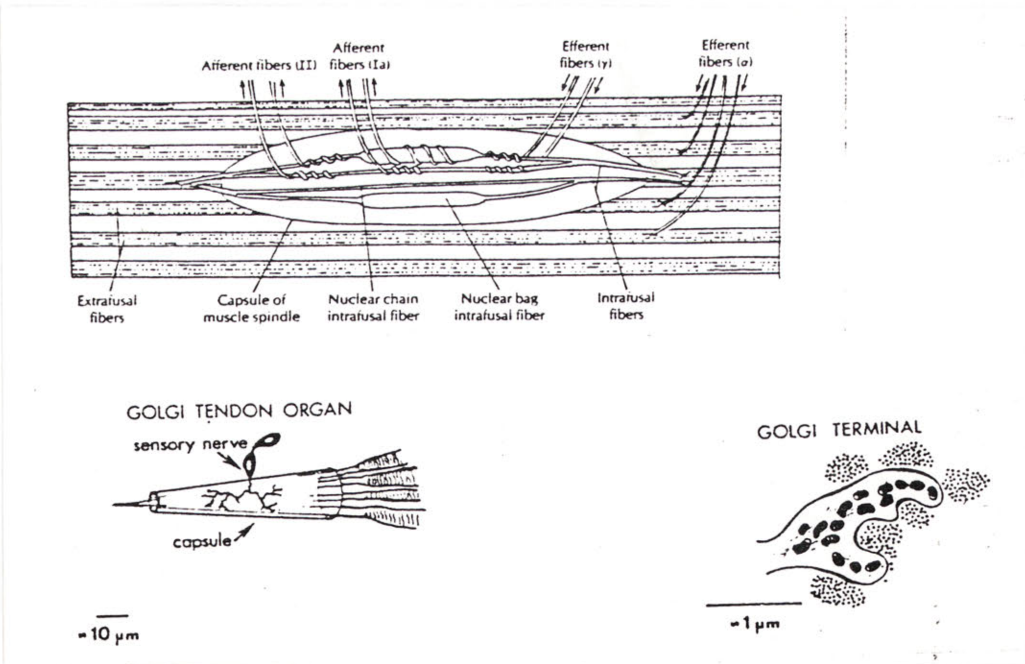


Figure 38 Muscle spindle (A), and Golgi tendon organ (B) (Bridgman, 1968).

The extrafusal muscle fibres receive innervation by way of the alpha efferent motor neurons. Most of these have their cell bodies in the trigeminal motor nucleus. Stimulation of these neurons therefore causes the group of extrafusal muscle fibres innervated by this nucleus to contract.

From a functional standpoint the muscle spindle acts as a length monitoring system. It constantly feeds back information to the central nervous system regarding the state of elongation or contraction of the muscle. When a muscle is suddenly stretched, both its extrafusal and its intrafusal fibres elongate.The stretch of the spindle causes firing of the Group I and Group II afferent nerve endings leading back to the central nervous system. When the - efferent motor neurons are stimulated the extrafusal fibres of the muscles contract and the spindle is shortened. This shortening brings about a decrease in the afferent output of the spindle. A total shut-down of the spindle activity would occur during muscle contraction if there were no gamma efferent system. Stimulation of the gamma efferents causes the intrafusal fibres of the muscle spindle to contract. This can elicit afferent activity from the spindle even when the muscle is contracting. Efferent drive can therefore assist in maintaining muscle contraction. It is believed that the gamma-efferent system acts as a mechanism to sensitize the muscle spindles. Thus the fusimotor system can be considered to be acting as a biasing mechanism that alters the firing of the muscle spindle. It should be noted that the gamma efferent mechanism is not as well investigated in the masticatory system as in spinal cord systems. Although it appears to be active in most of the masticatory muscles, some apparently have no gamma efferents.

All jaw elevator muscles in man have been found by Freimann (1954), Gill (1971), and Honee (1966) to contain muscle spindles. For a long time there was doubt regarding the presence of spindles in the human lateral pterygoid muscle but Honee (1966) and Gill (1971) has shown that they exist.

The elevator muscles contain many muscle spindles and should be classed between larger limb muscles containing relatively few spindles, and muscles responsible for fine finger movements. These muscles also occupy an intermediate position in terms of their speed of contraction (Matthews, 1975).

By contrast, the jaw depressor muscles are not found consistently to contain spindles (Voss, 1956; Bossy, 1958).

There is good evidence in the literature that muscle spindles are not as randomly distributed in jaw muscles as they appear to be in many larger limb muscles.

The masseter muscle in man and animals appears to have the highes concentration of spindles in its deep portion (Karlsen, 1965; Voss, 1956; Kubota and Masegi 1972).

The medial pterygoid in animals is reported to have most of its spindles concentrated to its medial portion (Karlsen, 1965; Kubota and Masegi 1972).

In the lateral pterygoid in man the spindles appear to be present in the mid-portion (Honee, 1966).

Muscle spindles in different parts of a muscle give different responses to a given stimulus (Henatsch et al. 1975). The apparent concentration of spindles in certain locations of the elevator muscles implies that the central nervous system uses information from those parts preferentially.

5.3.2. Golgi Tendon Organs

The tendon organs named after Golgi surround the collagen bundles which are coupled in series with the extrafusal muscle fibres at the end of the muscle.

Szentagothi (1948) and Touloumis et al. (1975) observed Golgi tendon organs in the cat masseter muscle.

Each of these Golgi tendon organs consists of tendinous fibres surrounded by lymph spaces enclosed within a fibrous capsule. Afferent fibres enter near the middle of the organ and spread out over the extent of the fibres (Merrillees, 1962; Bridgman 1968).

Tension on the tendon stimulates the receptors in the Golgi tendon organ. Therefore contraction of the muscle also stimulates the organ. Likewise, an overall stretching of the muscle creates tension in the tendon and stimulates the organ.

At one time it was thought that the Golgi tendon organs had a much higher threshold than the muscle spindles and therefore functioned solely to protect the muscle from excessive or damaging tension. It now appears that they are more sensitive and are active in reflex regulation during normal function.

The Golgi tendon organs primarily monitor tension whereas the muscle spindles primarily monitor muscle length.

5.3.3. Pacinian Corpuscles:

The pacinian corpuscles are large oval organs made up of concentric lemellae of connective tissue. At the centre of each corpuscle is a core containing the termination of a nerve fibre. These corpuscles are found in the tendons, joints, periosteum, tendinous insertions, fasci, and subcutaneous tissue (Granit, 1955,1970, Kawamura et al., 1967).

Pressure applied to such tissues deforms the organ and stimulates the nerve fibre. There is a wide distribution of these organs, and because of their frequent location in the joint structures they are considered to serve principally for the perception of movement and firm pressure.

5.3.4. Nociceptors.

Generally nociceptors are sensory receptors that are stimulated by iritation and transmit this information to the central nervous system by way of the afferent nerve fibres.

Nociceptors are located throughout most of the tissues in the masticatory system. There are several general types: some respond exclusively to mechanical and thermal stimuli, others respond to a wide range of stimuli, from tactile sensations to noxious injury, still others are low-threshold receptors specific for light touch, pressure or facial hair movement. The last type is sometimes called mechanoreceptors.

The nociceptors primary function is to monitor the condition, position, and movement of the tissues in the masticatory system. When conditions exist that either are potentially harmful or actually cause injury to the tissue, the nociceptors relay this information to the central nervous system as sensations of discomfort or pain.

5.4 Neuromuscular Function.

In 1975 Storey was told of a student project conducted under Posselt which demonstrated that infiltration of the temporomandibular joint with local anaesthetic resulted in a change in postural position.

Schaerer et al. (1966) published a brief paper on two patients in which the effect of anaesthesis of the gingiva, periodontium and temporomandibular joints was investigated on chewing movements. The anaesthesia reduced the patient's ability to control the food bolus but the performance of chewing movements was not affected.

Lowe (1975) investigating the genioglossus and mylohyoid activity during various opening angles found that the EMG activities of these to muscles started at 2^{o} opening and was quite marked at 42^{o} and 63^{o} of opening, but infiltration with local anaesthetic into both temporomandibular joint capsules abolished the response completely.

Kawamura et al. (1974) during their investigations on the role of sensory information observed three young patients who had fractures of condylar heads which were ankylosed in the fossal and the patients were supplied with pseudo-joints on both sides. According to Kawamura they could chew very well without the sensory information from the joints.

Looking to clinical examples we can conclude that all these sensory receptors provide constant feedback to the central nervous system. This input is continually monitored and evaluated day and night, and during activity and relaxation.

The central nervous system evaluates and organizes the sensory input and initiates appropriate efferent input to create a desired motor function.

5.4.1 Reflex Action

A reflex action is the response resulting from a stimulus that passes as an impulse along an afferent neuron to a posterior nerve root or its cranial equivalent, where it is transmitted to an efferent neuron leading back to the skeletal muscle. A reflex action maybe monosynaptic or polysynaptic. In a monosynaptic reflex the afferent fibre directly stimulates the efferent fibre in the central nervous system while in a polysynaptic reflex the afferent neuron stimulates one or more interneurons in the central nervous system, which in turn stimulate the efferent nerve fibres.

5.4.1.1 The Myostatic Reflex

The myostatic or stretch reflex is the only monosynaptic jaw reflex. When a skeletal muscle is quickly stretched, this protective reflex is elicited and brings about a contraction of the stretched muscle.

Clinically this reflex can be demonstrated by relaxing the jaw muscles, allowing the teeth to separate slightly. A sudden downward tap on the chin will cause the muscle spindles within the masseter suddenly stretch, afferent nerve activity is generated from the spindles. These afferent impulses pass into the brainstem to the trigeminal motor nucleus by way of the trigeminal mesencephalic nucleus, where the primary afferent cell bodies are located. These same afferent fibres synapse with the alpha efferent motor neurons leading directly back to the extrafusal fibres of the masseter. Stimulation of the alpha efferent by the Ia afferent fibres causes the muscle to contract. This will cause the jaw to be reflexly elevated and the constraction of the masseter will result in tooth contact.

The myostatic reflex occurs without specific response from the brain and is very important in determining the resting position of the mandible. The myostatic reflex is a principal determinant of muscle tonus in the elevator muscles. As gravity pulls down on the mandible, the elevator muscles are passively stretched, which also creates stretching the muscle spindles. This information is reflexly passed from the afferent neurons originating in the spindles to the alpha motor neurons that lead back to the extrafusal fibres of the elevator muscles. Thus passive stretching causes a reactive contraction that relieves the stretch on the muscle spindle.

The myostatic reflex and resulting muscle tonus can be influenced by the higher centres via the fusimotor system. The higher centres bring about increased gamma efferent activity to the intrafusal fibres of the spindle. As this activity increases, the intrafusal fibres contract, causing a partial stretching of the nuclear bag and nuclear chain areas of the spindles. This lessens the amount of stretch needed in the overall muscle before the spindle afferent activity is elicited. Therefore the higher centres can use the fusimotor system to alter the sensitivity of the muscle spindles to stretch.

When a muscle contracts, the muscle spindles are shortened, which causes the afferent activity output of those spindles to shut down. If the electrical potential of the afferent nerve activity is monitored, a silent period will be noted during this contraction stage. Gamma efferent activity can influence the length of the silent period. High gamma efferent activity causes contraction of the intrafusal fibres, which lessens the time the spindle is shut down during a muscle contraction. Decreased gamma efferent activity lengthens this silent period.

The silent period of the afferent nerve activity should not be confused with the masticatory muscle silent period. The masseteric silent period has been investigated by McNamara (1976), Williamson (1982), Skiba and Laskin (1981), Bassette and Snatkin (1979), Hellsing and Klineberg (1983) by monitoring the EMG activity of the muscles of the mastication.

It has been suggested by McNamara (1976) that during the clenching a sudden downward tap to the chin activates the muscle spindles, which causes the information to be relayed to the central nervous system by way of the mesencephalic nucleus. An interruption then occurrs at the motor nucleus of cranial nerve V and no motor impulses are sent to the muscles of mastication for a very short time. The motor impulses then return and the muscle continues to contract. The time of no electrical activity is called the silent period.

It has been suggested by Skiba (1981) and Bassette (1971) that the silent periods of patients with functional disturbances of the masticatory muscles are significantly increased.

5.4.1.2. Nociceptive (flexor) Reflex

The nociceptive or flexor reflex is a polysynaptic reflex to noxious stimuli and therefore is considered to be protective. In the masticatory system this reflex becomes active when a hard object is suddenly encountered during mastication.

As the tooth is forced down on the hard object, a noxious stimulus is received by the tooth and surrounding periodontal structures. The associated sensory receptors trigger afferent nerve fibres, which carry the information to the interneurons in the trigeminal motor nucleus. The action taken during this reflex is more complicated than the myotatic reflex in that the activity of several muscle groups must be co-ordinated to carry out the desired motor response. not only must the elevator muscles be inhibited to prevent further jaw closure on the hard object, but the jaw opening muscles must be activated to bring the teeth away from potential damage.

As the afferent information from the sensory receptors reaches the interneurons, two distinct actions are taken. Excitatory interneurons leading to the efferent fibres of the jaw opening muscles are stimulated. This action causes these muscles to contract.

At the same time the afferent fibres stimulate inhibitory interneurons, which have their effect on the jaw elevating muscles and cause them to relax. The overall result is that, the jaw quickly drops and the teeth are pulled away from the object causing the noxious stimulus. This process is called antagonistic inhibition, and it occurs in many reflex actions throughout the body.

5.4.2 Antagonistic Action

The control of antagonistic muscles is of vital importance in reflex activity. There are certain groups of muscles that primarily elevate the mandible as well as others that primarily depress it. For the mandible to be elevated by the temporalis, medial pterygoid, or masseter, the suprahyoid muscles must relax and lengthen. Likewise, for it to be depressed, the suprahyoids must contract while the elevator muscles relax and lengthen. The neurologic controlling mechanism for these antagonistic groups is known as reciprocal innervation.

This phenomenon enables smooth and exact control of mandibular movement to be achieved. For the skeletal relationship of the skull, mandible, and neck to be maintained, each of the antagonistic muscle groups must remain in a constant state of mild tonus. This will overcome the skeletal imbalances caused by gravity and keep the head in what is termed the postural position.

Muscles that are in full contraction, fatigue rapidly because of the decreased blood flow and eventual build up of metabolic by products in the muscle tissues. By contrast, muscles in tonic contraction allow proper blood flow to bring needed metabolic products to the muscle tissues. Therefore normal muscle tonus does not create fatigue.

CHAPTER VI

BIOMECHANICS OF THE MANDIBULAR MOVEMENT

The mandible is a bone that is joined to the skull by ligaments and suspended in a muscular sling. When the elevator muscles are called upon to function, their contraction raises the mandible, so contact is made and force is applied to the skull in three areas: the two temporomandibular joints and the teeth. Since these muscles have the capability of providing heavy forces, there is a great potential for damage to occur at the three sites. Thus there is a need to examine these areas closely to determine the optimum anatomic relationship that could prevent, minimize, or eliminate any breakdown or trauma. The fact that not only strong muscles but two temporomandibular joints are also connected to the same bone further complicates the biomechanical analysis of the entire masticatory system. Each of the joints can act simultaneously but differently and yet not completely without influence from the other. A sound understanding of the biomechanics of the temporomandibular joint is essential and basic to the study of function and dysfunction in the masticatory system.

The biomechanics of the movement of the mandible is complex because its widely separated temporomandibular joints allow it to translate in any direction to a limited extend and/or rotate about any axis in space.

Due to translational and/or rotational movements the biomechanics of mandible have proved difficult to analyze. Because of the complexity of the system, biomechanical models of the mandible have usually been restricted to analysing forces and/or movements in a single plane (Barbanel, 1972, 1974; Grant 1973a, 1973b; Pruim et al. 1978, 1980; Throckmorton et al. 1985a, 1985b). The predictions made from such models proved to be incomplete because the forces and movements operate in three dimensions and their projections onto a plane suppress what may be important biomechanical aspects of the system. Three dimensional models using finite element (Gupta et al., 1973; Norton et al., 1974; Knoell, 1977) and computerized model based on the principle of linear programming (Osborn and Baragar, 1985; Baragar and Osborn 1984) also has been constructed.

Even from the very early pioneering work of Ulrich (1896) there has been considerable argument about the axis of the opening movement, and about the critical problem of whether or not the joints are loaded when the teeth biting on an object (Hylander, 1975). Only recently (Hylander, 1979; Brehnan et al. 1981) have developed techniques to measure joint loads but this could only be used in animals and the results would be questionable.

Perhaps the most frequently used biomechanical model for the mandible has been the Class III lever, in which the condyle acts as a fulcrum, the masticatory muscles as applied force, and the bite pressure as resistance (Fig.39).

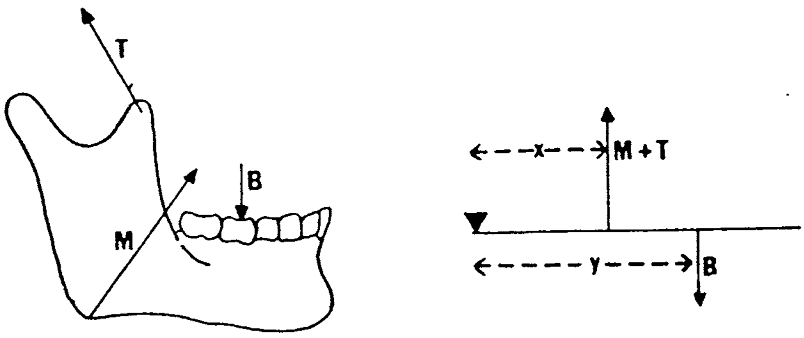


Figure 39 The mandible as a Class III lever.

For mechanical equilibrium, the Class III lever requires that the bite force be less than the applied force, so that the masticatory "machine" would have a mechanical advantage equal to, less than one. Some have objected that such an inefficient system is inherently improbable (Tattersall, 1973). Others have argued that the condyle and articular fossa are simply not structured to withstand the large resultant forces required by the model (Robinson, 1946; Roberts and Tattersall, 1974). Nevertheless, as Barbanel (1974) has emphasized, this type of histological interpretation is highly subjective, and many workers (Boucher, 1962; Noble, 1973; Hylander, 1975) have explained that the region is well-adapted to withstand stress.

Several other alternative biomechanical models also has been proposed. Hylander (1975) reviewed much of the literature on non-lever models of Wilson, 1920, 1921; Robinson, 1946; Frankel and Burnstein, 1970; Gingerich, 1971 and Tattersall, 1973, and he concluded that on the basis of electromyographic and anatomical data, the non-lever models are largely invalid.

Few workers (Crampton and Hotten, 1967; Maglio, 1972) have recognized the biomechanical significance of the direction and the magnitude of the muscles of mastication and have used the weight and/or cross-sectional area of masticatory muscles as estimates of relative force magnitudes.

Roberts (1974) and Roberts and Tattersall (1974) have also been concerned with whether or not there is force at the condyle, rather than with the mechanism of force generation. Roberts (1974) suggests that since the masseter, temporalis and bite force vectors can form a closed triangle, there is no additional reaction force at the condyle (Fig.40).

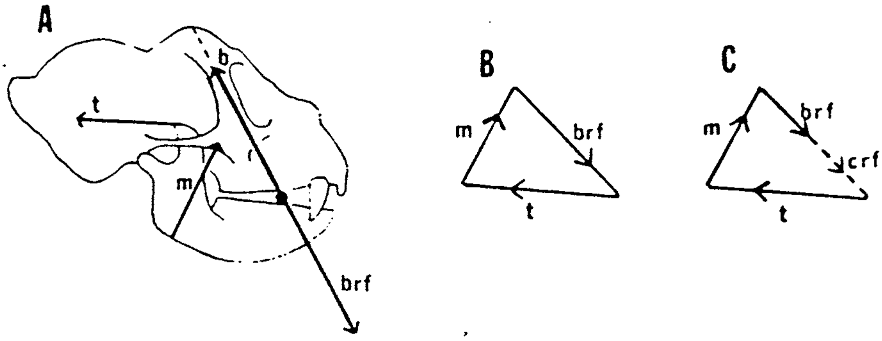


Figure 40 A and B demonstrate equilibrium between muscle forces and bite reaction force according to Roberts, (1974).

In this argument, each force is represented by a vector whose length is equal to the force magnitude. Since accurate data on the direction of the three forces, and certainly on the magnitude of the two muscle forces, are completely unknown, a triangle of these forces in equilibrium is entirely hypothetical.

Roberts and Tattersall (1974) state that their analysis assumes "no significant force is expended at the temporomandibular joint during elevation of the mandible". While many others do in fact suggest that large condylar reaction forces exist. (Roydhouse, 1955; Ostrom, 1964; Badoux, 1972; Hekneby, 1974; Barbanel, 1972, 1974; Hylander 1975, 1984; Walker, 1978; Pruim et al. 1978, 1980; Throckmorton, 1985).

Parrington (1934) has suggested that external forces on the mandible during mastication could be analyzed by considering the mandible a stationary beam, and several workers (Badoux, 1966; Boch, 1966; Bock and Kummer, 1968; Hylander, 1977) have examined the internal stresses and strains of mastication on the mandible with this approach. Only Badoux (1972), Walker, (1978) and Smith (1978) have clearly stated that mechanically the mandible can be considered to "change states" once contact is made.

Smith (1978) suggested that during mandibular elevation, the mandible acts as a Class III lever. However, when maximum masticatory forces are generated, probably during the working functional movement, it may be useful to model the mandible as a stationary beam (Smith, 1978) or structural girder (Boch and Kummer, 1968), (Fig.41).

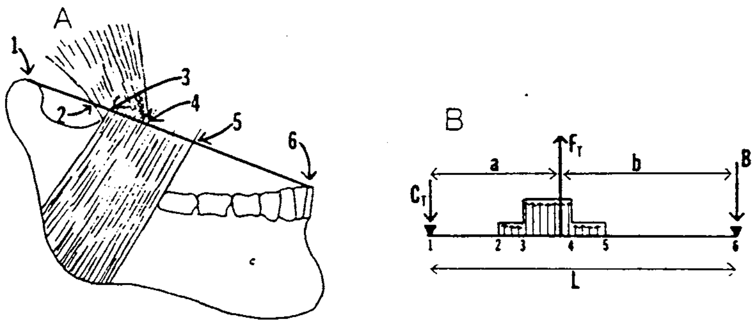


Figure 41 The mandible as a stationary beam (saggital projection, incisal biting), (Smith, 1978).

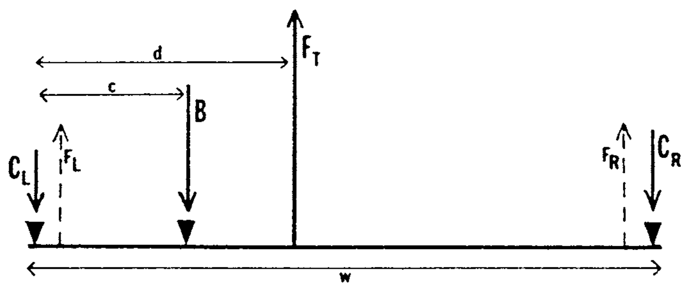


Figure 42 The mandible as a stationary beam frontal projection, unilateral molar biting (Smith, 1978).

6.1 Axis of the Opening Movements

Prosthodontic interest in the location of the axis of the mandibular movement appears to stem from practical considerations. The biomechanical need for an axis could be felt when attempting to simulate the natural movements of the mandible with mechanical devices.

Opinions regarding the axis of the opening movement of the mandible are quite abundant in the anatomical literature. The most extreme opinion is: "Opening of the mouth is a simple movement. The mandible, slung in muscle, rotates about an axis that passes through the mandibular foramina "(Last, 1954; Hamilton, 1956; Lockhard et al., 1959). While others speak of "a condylar hinge axis (Johnston and Whillis, 1954), an undefined hinge axis (Terry and Trother, 1953), a moving axis (Scott and Symons, 1958), or an instantaneous centre of rotation (Ulrich, 1896; Grant 1973, 1973b).

According to the glossary of Prosthodontic terms prepared by the Academy of Denture Prosthetics (1977), Axis is: "A straight line around which a body may rotate." Which is defined as three separate axes:

longitudinal axis : An imaginary anteroposterior line through a mandibular condyle around which the mandible may rotate in a rolling motion.

mandibular axis : (condylar axis, condyle axis, hinge axis, transverse axis)
Imaginary line through two mandibular condyles around which the mandible may rotate without translatory movement.

sagittal axis : An imaginary vertical line through a mandibular condyle around which a mandible may rotate in a yawing or pitching motion.

Prosthodontic history is somewhat incomplete as to the origins of the thesis of a transverse horizontal axis (Brekke, 1959). In its purest form the transverse axis is usually thought of as exhibiting a two dimensional effect and as being independent of the vertical and sagittal axes.

Some investigators have explored the degree of precision could be expected with an axis location procedure (Kurt and Feinstein, 1951; Borgh and Posselt, 1958; Lauritzen et al., 1961; Preston, 1979).

A major challenge to the traditional concept of a single "intercondylar" axis was postulated by Page (1960). He postulated the existence of two mutually independent, noncolinear axes or, simply, that each condyle had its own axis of rotation.

Trapozzano and Lazzari (1967) reported the location of multiple hinge axis points along the path of the translatory movement of the condyle.

The anatomic relation of the transverse horizontal axis has been explored by a number of investigators, and most have made statements regarding the relation of the axis to the condyle (Good, 1953; Weinberg, 1959).

No unaniminity of opinion exists, and the conclusions range from "the transverse axis passes through or near the condyles" (Weinberg 1959), "within the limites of accuracy imposed by individual operators, equipment, and patient variations, a single transverse horizontal axis can usually appear to be located" (Preston, 1979), to "we never found the axis located in the condyle" (Le Pera, 1964).

Most dentists with experience in hinge axis location admit that there are times when no hinge axis can be located (Gregory et al. 1969). Obviously from a clinical point of view the location of a transverse horizontal axis has many complexities but the kinematic location of a transverse horizontal axis has a definite biomechanical merit.

Various arguments also have been presented for the existence of mandibular foramina axis (Last, 1954; Moss, 1960). But Koski (1962) suggested that the opening axis could not pass through the foramina, as this position could lead the condyles against the eminences in the very beginning of the opening act.

Navarkari (1956) postulated that the opening axis was always outside the condyle, and in most instances, it was located in the region of the mastoid process when the mandible moved from rest to the occluding position.

Koski (1962) agreeing with these findings suggested that during the first phases of opening the axis might be most frequently located at the mastoid region.

Hjortsjo (1954), and Moss (1959) reported on the biaxial nature of the mandibular opening movement. According to Hjortsjo, one axis of the movement passes through the centres of the condyles and another through the centres of the articular eminences. Two simultaneous movements around these axes results in the opening of the mandible.

Anatomists have known for many years that most of the bones of the body do not rotate around fixed axes of rotation but rather around a moving instantaneous centre of rotation. The path of the instantaneous centre of rotation was first observed by Ulrich (1896), then Chissin (1906), Hall (1929) and Grant (1973a, 1973b). In 1974 Stern debated the relevance of the instantaneous centre of rotation and rejected its biomechanical significance.

The complexity of the axis of the movement of the mandible and diversity in opinions has also created different biomechanical arguments about the critical and apparently simple problem of whether or not the joints are loaded when biting on an object.

6.2 Non-Lever Action Hypotheses:

Various workers have suggested, either directly or indirectly, that there is little or no reactive force at either mandibular condyle. Wilson (1920, 1921), argued that because the resultant force of the temporalis, masseter, and medial pterygoid muscles lies perpendicular to the occlusal plane, there can be no reaction force at the mandibular condyle; and thus it was incorrect to view the human mandible as a lever. Wilson's argument was not correct; to eliminate reaction force at the temporomandibular joint, the resultant force must pass through the bite point, not simply be oriented perpendicular to the occlusal plane.

Robinson (1946), based his non-lever notion of mandibular function on essentially two assumptions. According to Robinson the resultant adductor muscle force passes through the first molar tooth. Assuming that the resultant force has been correctly determined, biting on the first molar in this projection would result in a non-lever action of the mandible. However, biting anterior or posterior to this tooth in this projection would result in lever action. Gingerich (1971) pointed out that the resultant force determined by Robinson was incorrect because the direction of the combined force of the temporalis and masseter was incorrectly determined.

Robinson also noted that immediately above the mandibular condyle, located along the articular disc, is a synovial layer. Since synovial tissue is never found in a stress bearing region (only adjacent to it), he argued that this region cannot be under compression and therefore, could not act as a fulcrum. In addition, Robinson cites the occurrence of the fibrocartilaginous articular disc as further evidence for this being a non-stress bearing joint. The presence of blood vessels, nerves and lymphatics in the articular disc, and the paper-thin bony roof of the mandibular fossa further substantiates his claim that this region is under negligible amounts of stress.

Moss (1960) noted that, the mandibular fossa is a "receptacle" for the head of the mandibular condyle when the jaws are closed. The actual stress bearing articular portion of the temporomandibular joint is located between the articular tubercle of the temporal bone and that part of the mandibular condyle immediately facing it. Therefore, whereas the paper-thin bony roof of the mandibular fossa is indeed indicative of a non-stress bearing region, the articular tubercle is composed of thick cancellous bone with a dense cortical plate (Sicher, 1960), and thus probably is capable of bearing considerable amounts of reactive force (Hylander, 1975).

According to Rees (1954), Griffin and Sharpe (1960), Sicher and Bhaskar (1972), the region of the articular disc situated between the mandibular condyle and the articular eminence, contains an avascular portion that lacks both nerves and a synovial layer.

Frank (1950) has also suggested non-lever action of the mandible. His arguments are based on a radiographic analysis of the human mandibular condyle. Frank noted that the condyle was never in direct contact with the articular eminence; he concluded that the condyle was not functioning as a fulcrum and therefore the mandible could not function as a lever.

Frankel and Burstein (1970) have also concluded that it is not necessary to have reaction forces acting on the mandibular condyle during biting. Although this conclusion could be true, it does not necessarily follow from their analysis because they have also incorrectly determined the resultant muscle force(fig.43).

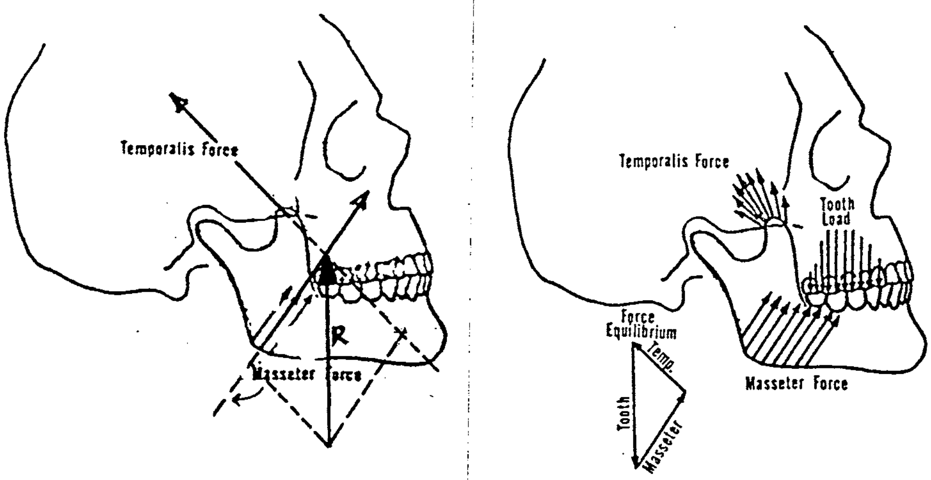


Figure 43 Comparison of Robinson (1946), (a) and Frankel and Burstein (1970), (b) models.

Another worker to suggest non-lever action in the human mandible is Gingerich (1971). He has stated that "During biting, the jaw is functionally a link between the adductor muscle force and the bite force rather than a lever". He also states that "...other adductor muscles are not aligned with any bite point; their force of contraction is divided between useful bite force and wasted reaction force at the jaw joint".

Thus, a more accurate statement of Gingerich's view on mandibular function would be that the jaw is both a link and a lever, despite his statements to the contrary in the same paper.

Tattersall (1973) has also suggested that the mandible does not function as a lever during chewing or biting. He pointed out that the morphology of the jaw is incapable of dissipating considerable stress". The condylar neck is sufficiently weak for it to be said categorically that it could not have withstood the forces that would have been imposed upon it in a lever system". Tattersalls work on mandibular function are based on an analysis of the masticatory apparatus of the Archaeolemurinal, now an extinct group of Malagasy lemurs. If the mandible functions as a lever, the condyle and its neck must be stressed maximally during powerful incisal biting, when reaction forces will probably be greatest. In this situation, the condylar region will be sheared or bent.

The ability of this region to resist shearing stress is largely a function of its cross sectional area. During bending its strength is a function of the second moment of area of the cross section (Alexander, 1968). In order to test the hypothesis that the condylar neck is too weak to withstand reaction forces during biting Hylander (1975) analyzed the strength of the condyle region of the human mandible(Fig.44). He found that the condylar neck can be capable of withstanding an average shearing stress of up to 1745 N (178 kg). He also concluded that reaction forces of this magnitude could not be present during powerful biting.

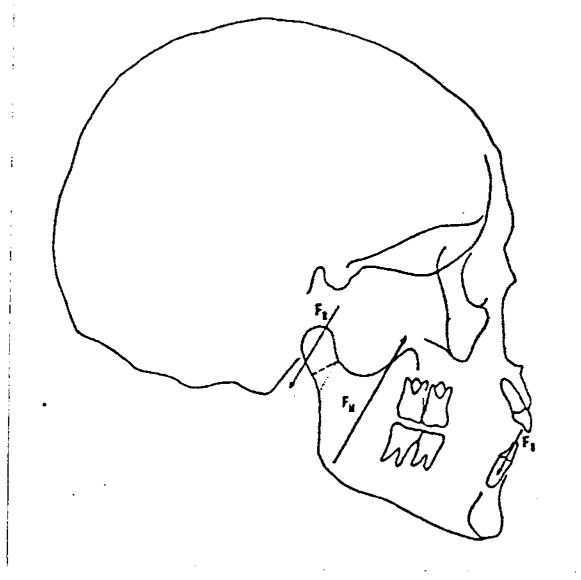


Figure 44 Forces acting on the mandible during incisal biting (Hylander, 1975).

6.3 Lever Action Hypothesis:

Another approach to the analysis of the mammalian mandible, and in particular the human mandible, has been the assumption that it acts as a lever. Most workers have agreed with this assumption and have usually analyzed the mammalian jaw in the lateral projection under static conditions and have calculated moment arms for both the bite and muscle forces about the mandibular condyle. However, Moss (1960) and Grant (1973) considered this approach to be fundamentally incorrect. These workers have positioned the centre of rotation in the human mandible along the region of the mandibular foramen (Moss 1960), or is variable depending on the position of the mandible (Grant 1973, Lupkiewicz et al., 1982). Actually it has been shown that the instantaneous centre of rotation in the lateral projection is also dependent on whether the balancing or working condyle is involved; on the texture of the food; and on individual variation (Gibbs et al. 1969).

Stern (1974) suggest that Grant's claims about the biomechanical significance of the instantaneous centre of rotation would not have been made, had Grant taken into account the moments for the bite and reaction forces, as well as those of the masticatory muscle forces.

Most workers who adhere to lever action of the mandible have taken moments about the mandibular condyle in the lateral projection. As noted by Stern (1974) this approach is appropriate and is performed in two dimensional analysis in order to eliminate the necessity of taking reaction force moments into account.

Although analyzing the mammalian jaws solely in the lateral projection is probably appropriate for incisor or bilateral molar biting, such an analysis is incomplete for unilateral biting.This is because the projected bite point is never actually located in the mid-sagittal plane.Therefore if three dimensional analysis not possible it would be also useful to analyze the human mandible in the frontal projection (Hylander, 1975),(Fig.45).

Such an analysis were carried out by various investigators but these workers only considered the muscle force on the working side of the jaw during unilateral molar biting. This is a very serious mistake since it has been conclusively demonstrated by Moller (1966) that the mandibular adductors on both sides are electrically active during unilateral molar biting.

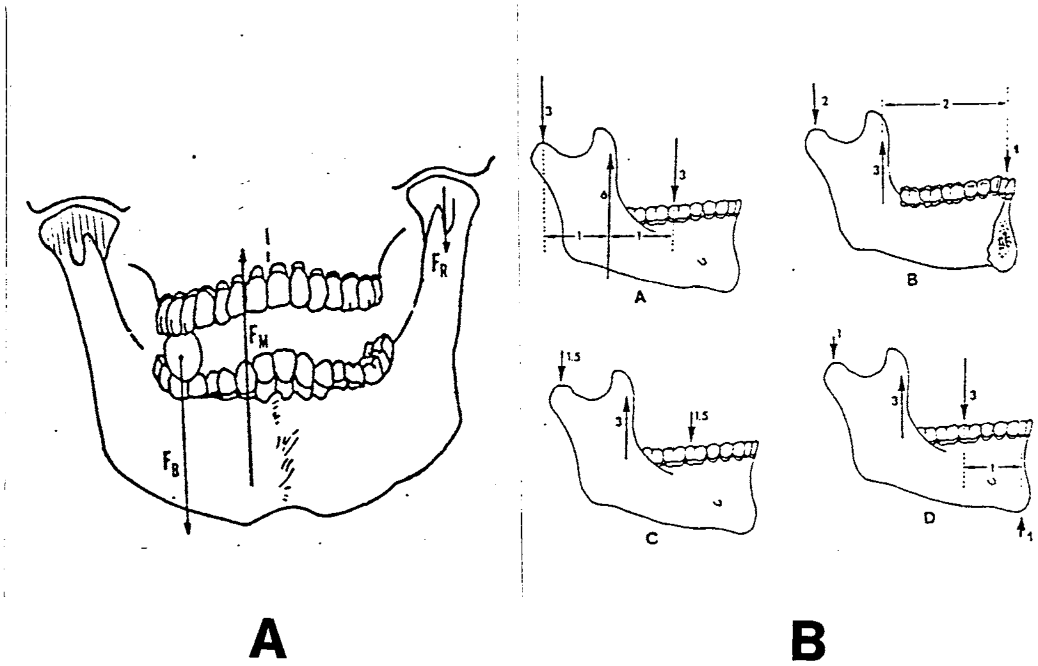


Figure 45 Forces acting on the mandible in the frontal projection during unilateral molar biting (Hylander 1975) (A), and biomechanics of the mandible in the lateral proection (Hylander, 1979) (B).

The electromyographic data suggest that the muscles on the biting side are slightly more active than on the non-biting side (Moller 1966, 1970). Hylander (1979) and Gysi (1921) postulated that during powerful unilateral biting forces acting on the balancing condyle are probably greater than those on the working condyle. This probably explains why individuals with a painful temporomandibular joint bite on the side of the diseased joint (Ramjford and Ash, 1971). Biting on this side possibly results in less pain because the working condyle has less reactive force acting on it. Similarly incisor biting could be more painful to the diseased joint than molar biting because the reaction forces acting on the condyles are maximal. Hylander (1979) has implanted strain gauges in the condylar necks of macaques and found 50-60% less strain on the working side condyle compared to the balancing side when an animal was chewing apples.

According to Hylander (1979), the contralateral mandibular condyle could have a large compressive reaction force acting across it during both unilateral mastication and molar biting, while lower levels of compressive stress act across the ipsilateral condyle during mastication. In some instances during unilateral molar biting, the ipsilateral condyle might even be free of stress, or there may be tensile stress acting across it. He further suggests that if the articular surfaces of the temporomandibular joint tend to be separated due to tensile stress, then possibly one of the functions of the temporomandibular ligament is to prevent the ipsilateral mandibular condyle being moved too far off the articular eminence.

Widening or narrowing of the dental arches may not, as Hylander (1975, 1984) has suggested, change the total magnitude of force at the condyles, but only the proportion borne by each condyle. The nature of muscle positions in the sagittal plane dictates that condylar forces will occur. If large condylar forces are present, the most efficient mechanism might be an equal distribution between right and left sides, minimizing the maximum load applied to each. This can be achieved by either widening the bicondylar dimension or narrowing dental arch width. Using a lever model Hylander (1975, 1979) reached the same conclusion regarding the pattern of morphological change to be expected but for different reasons.

In most of the early biomechanical models, the forces exerted by ligaments, soft tissues and other muscles were not considered, and the direction of the bite force and the direction and magnitude of muscle forces were largely unknown. It was assumed, that the mandible is made of homogeneous and isotropic rigid material with uniform cross-section and with all loads and reactions in the same plane.

Nevertheless, the many functional analysis of muscle positions using the lever model have demonstrated that important insights can be gained from this type of simplified approach.

CHAPTER VII

BIOMECHANICAL ANALYSIS OF THE MANDIBLE

Analytical calculation of internal forces in the joints and muscles of the musculoskeletal system has been one of the challenging topics in the field of biomechanics.

During function there are forces present at the skeletal joints. A joint is the interface where relative motion is permitted between two rigid bodies. It is the geometry of the contacting surfaces which defines the relative motion. In an anatomical joint, the articular surfaces are usually geometrically complex and somewhat compliant so that all anatomical joints are in reality six-degree-of freedom joints. This means that six independent parameters must be measured and described if the position of one body segment is to be defined accurately relative to the other. Unfortunately, both the measurement and the description of six-degree-of freedom motion are difficult to accomplish. In fact, it has only been in the last three decades that transducers and the electronic support equipment have become accurate enough to measure six-degree-of freedom anatomical motion reliably.

In general, equilibrium equations for force analysis of the musculoskeletal system are formulated by free-body force analysis in which the musculoskeletal system is considered as a linkage system of intercalated bony segments balanced by two groups of forces. The first group consists of the resultant passive constraint forces and moments exerted by the joint articulating surfaces and ligaments. The other group consists of the active tensions of tendons and muscles.

In the force equilibrium equations, the "coefficients" of the muscles represent the components of the unit force vector of the muscles in the respective co-ordinate axes. In moment equilibrium equations, the coefficients of the muscles represent the three components of the moment arm of the muscle force vector about the joint centre with respect to the three axes.

In the analytic determination of these muscle and joint forces, a frequently encountered difficulty is that the problem formulated is an indeterminate one, in which the number of unknown variables exceeds that of the equations. Therefore, no unique solution can be obtained.

Many approaches have been developed to resolve this indeterminate problem. The most commonly used method is reduction of the excess number of unknown variables, either by grouping functionally similar muscles (Paul, 1965, 1967; Morrison, 1967; Nicol et al. 1977; Berme et al. 1977), or by eliminating them based on electromyographic observations (Barbanel, 1974; Pruim et al. 1978, 1980).

For example in the elbow, forces shared between co-operating muscles had been based on arbitrary assumed ratios (Simpson, 1975; Nicol, 1977), or further equations were introduced based on the hypothesis that force distribution among muscles depends on their cross sectional area during maximal forceful activities (Pauwels, 1980; Amis et al. 1980; Hui et al. 1978).

Another method of solution is to restrict the number of non-zero joint and muscle forces to the number of available equations and permute the choice of these non-zero variables from the set of all the muscle and joint forces. The technique has been used by Chao et al. (1976) for the finger, and Nissan (1981) for the knee, but has the disadvantage that it produces a range of possible values, not a unique solution.

Alternatively, the optimization methods has also been used for a unique solution. By using the optimization method, not only a unique solution can be obtained, but possible physiologically or pathologically based rationales for the solution can be associated. Based on the hypothesis of energy conservation, summation of muscle forces was initially selected as the optimization criterion (Yao, 1976; Crowninshield, 1978; Arvikar et al. 1978). However, the solution so obtained did not agree well with the evidence from electromyographic experiments. Weighted, linear combinations of muscle and joint forces have thus been used as new criteria. Unfortunately, due to the inherent characteristics of the linear optimization approach, only a few active muscles were predicted simultaneously for each solution. In light of this limitation, a nonlinear optimization method using nonlinear combinations of unknown variables as the objective function was used for solution. By comparing the analytic results with the number of electromyographically active muscles, the sum of squares of muscle stress was found to be the most feasible criterion (Crowninshield et al. 1981). Unfortunately, mathematically the nonlinear optimization procedure is complicated, and global convergence of solution is not always guaranteed. In 1972 linear programming algorithm was used to obtain the solution more efficiently; (Barbanel, 1972; Campbell, 1977; An et al. 1984; Osborn and Baragar 1985). The solution of muscle force distribution based on the proposed predicted the same number of active muscles for the same given loading condition as that using the sum of square of muscle forces or muscle stress as the objective function.

However, the concept of this optimization approach was quite different from those previously used in summations of muscle force or stress, or their nonlinear combinations. All optimization procedures previously used to minimize the sum of unknown force , variables have been more or less based on consideration of overall efforts of the system. However, from an energy storage and transport viewpoint, each muscle bundle has its own storage and blood supply. An et al. (1984) in constructing the optimization criteria assumed that for each muscle, the endurance was inversely proportional to the muscle stress. They have reported that the method was well suited for two and three dimensional analyses.

The biomechanics of the mandible have proved difficult to analyze due to the complexity of movements. Even though estimation of the temporomandibular forces from mathematical models has a long history it has lead to quite conflicting results. The greatest controversy - as mentioned in previous chapter in detail - has been over whether the temporomandibular joint is even a load bearing joint or not. It is however now generally accepted that the temporomandibular joint is load bearing under most normal conditions, and that the balancing side condyle is more heavily loaded during a unilateral bite.

A complete model of all the forces contributing to the temporomandibular joint reaction force would be complex, including at least ten muscle force vectors, the bite force vector, and force vectors contributed by connective tissues, with all vectors determined in three dimensions.

With an exception of few previous investigations and the present study, most of the temporomandibular joint force models are two-dimensional, two/four muscle models. The predictions made from such models are incomplete because the forces and moments operate in three dimensions and their projections onto a plane suppress what may be important biomechanical aspects of the system.

In general terms, calculations of the temporomandibular joint reaction force requires three sets of information:

i) the magnitude and direction of the bite force,

ii) the magnitude and directions of each muscle force, and

iii) the lengths of the moment arms of the bite and muscle forces.

The magnitude and direction of the bite force can be directly measured with force transducers (Hylander, 1978 ; Pruim et al., 1978) and the lengths of the moment arms can be estimated from lateral cephalograms of human subjects.

However, as in the case for measuring joint forces directly, direct measurement of the magnitude and direction of muscle forces requires invasive techniques not suitable for human subjects and probably the results would be questionable (Osborn and Baragar, 1985).

7.1 Direction of Muscle Forces.

Little has been done to accurately determine directions of force for the jaw muscles; indeed, there have been no experimental studies to accurately measure the directions of the jaw muscle forces. A number of studies arbitrarily selected directions for the jaw elevator forces which were consistent with preconceived biomechanical theories but without any supporting biological data (Wilson, 1920; Robinson, 1946; Page, 1954; Gingerich, 1971; Roberts, 1974; Smith, 1978).

Several biomechanical studies have used a single resultant muscle force, but no methods were given for determining this resultant (Hekneby, 1974; Gosen, 1974; Greaves, 1978).

A relatively simple method for estimating the directions of the jaw muscle forces is to use the line between the centres of the origins and insertions of the muscles (Mainland and Hiltz, 1934; Jansen and Davy, 1975; Pruim et al. 1980; Weijs and Dantuma, 1981; Osborn and Baragar, 1985); or to use prominant landmarks in lateral cephalograms (Throckmorton et al., 1980; Finn et al. 1980a, 1980b).

More complicated attempts to determine the direction of the jaw muscle forces have taken into account relative strengths of different portions of the muscles. Gysi (1921) divided the jaw muscles into arbitrary portions, determined the direction of the vector bisecting each portion, and calculated the magnitude of each vector based upon the thickness of each portion. Carlsoo (1952) undertook a complex method of estimating force directions by estimating the cross-sectional area of portions of the muscle and taking into account the lengths of the muscle fibres and the pinnation angle in each portion.

The most complex analysis of jaw muscle force directions have used anatomical dissections to divide the muscles into as many as twelve different fascicles, and thus twelve different force vectors on each side (Gaspard et al., 1973a, 1973b, 1973c, 1974; Baron, 1975; Baron and Debussy, 1979).

Several investigators (Shumacher, 1961; Weijs, 1980) have argued that the direction of the jaw muscle forces is probably continuously changing during normal function and that assigning a single, constant direction for each of the jaw muscle forces is probably impossible. However, one should keep in mind that, the relationships between electromyographic activity and muscle force direction in different portions of the muscle, the cross sectional area of portions of the muscle, and changes in the relative positions of the origins and insertions of the muscles during function, has not been established experimentally.

7.2 Magnitude of Muscle Forces.

Two methods have been used to estimate the magnitude of the muscle forces. Gysi (1921), Mainland and Hiltz (1934), Carlsoo (1952), Schumacher (1961), Pruim et al. (1980) attempted to estimate the force generated by each muscle from the total cross sectional area of the muscle. Although techniques for measurement of the physiological cross-sectional area of complex muscles are improving, they still contain uncertainities (Weijs and Hillen, 1984). In addition it is highly possible that some parts of the muscles are selectively co-activated during normal functional movements.

In the second method (Barbanel, 1974; Pruim et al. 1978, 1980) the integrated electromyogram (IEMG) from each muscle was used as an estimate of muscle force. The IEMG value has been shown to be proportional to isometric tension (Inman et al., 1952; Lippold, 1952) but transformation of IEMG values into units of force requires an empirically determined proportionality constant which may not be the same for all muscles. Because of these uncertainties in determining the magnitude of the muscle forces, calculations of the joint reaction forces from mechanical models has been thought to be too imprecise for quantitative analysis of joint load.

7.3 Bite Force Analysis.

Bite force measurements have been used as a noninvasive method for assessing properties of the craniofacial complex including craniofacial biomechanics (Fields et al. 1982; Proffit et al. 1983; Ringqvist, 1973; Throckmorton et al., 1980, 1985) and the strength (Black, 1895; Brekhaus et al. 1941; DeBoever et al., 1978; Dechow and Carlson, 1982; Pruim et al. 1978, 1980), electrical activity (Garrett et al., 1964; Palla and Ash, 1981) and length-tension relationship (Dechow and Carlson, 1982, 1983; Manns et al. 1979; Nordstrom and Yemm, 1974; Thexton and Hiiemae, 1975) of the muscles of mastication.

Many studies have been carried out with the object of calculating the bite force. General reviews of investigations in this area are given by Brawley and Sedwich (1940), Strenger (1949), and Jenkins (1966).

There is a wide range of maximum values for bite force reported by different authors. Disparities in the materials investigated and differences in the techniques and instruments used may account for this.

Large inter individual variations in bite force were found in most studies, but few attempts have been made to explain these variations. In one study (Brekhaus et al., 1941) it was reported that females maximum biting loads range from 351N to 440N whereas males biting loads vary from 525N to 630N. The greatest maximum biting force reported is in the male Eskimo 1557N (Waugh, 1937).

Linderholm and Wennstrom (1970) reported that no correlation between bite force and muscle force or body build could be observed. Body build and muscle force by the extremities and of the trunk were positively correlated by Lindegard (1953), Asmussen and Heeboll-Nielsen (1961).

Ringqvist (1973) found that the muscle activity during maximal bite is positively correlated to mandibular prognatism, anterior inclination of the mandible and a small gonion angel. There were no significant relationship between bite force and maximal mandibular movements. They have concluded that in healthy individuals bite force and mobility of the mandible seems to function independent of each other.

The amount of force placed on the teeth during mastication varies greatly from individual to individual. A study by Gibbs et al. (1981) reports that the grinding phase of the closure stroke averaged 260N on the posterior teeth.

This represented 36.2% of the subject's maximum bite force. The maximum biting force during clenching ranged from 245N to 1245N. They also reported that there was no correlation of biting force either to sex or age.

It has also been reported that the maximum amount of force applied to a molar is usually several times that which can be applied to an incisor (Waugh, 1937).

In another study (Howell and Manly, 1948), the range of maximum force applied to the first molar was 405N to 880N whereas the maximum force applied to the central incisors was 129N to 227N.

Garner and Kotwal, (1973) has reported that the maximum biting force appears to increase with age up to adolescence.

It has also been demonstrated that individuals can increase their maximum biting force over time with practice and exercise (Brekhaus et al. 1941; Worner and Anderson, 1944).

Manns and Spreng (1977) in a study comparing submaximal masticatory force and EMG during prolonged isometric contractions of the human masseter and temporal muscles at different elongations, have found that around 12 to 20 mm jaw opening the masseter muscle reaches almost maximum elongation. It is generally accepted that muscular elongation, or more precisely the sarcomere length of each fibre of a skeletal muscle, determines its ability to produce active tension (Fulton, 1955). The tension created by contraction heightens as muscular elongation increases until it reaches a peak, and then tension decreases. Their results differ from those of Boos (1940) and Tueller (1969) which found that the strongest bite force was reached when jaw opening was near occlusion.

Boucher et al., (1959) found that there was progressive increase of masticatory force as jaw opening increased up to a level of 9 mm.

Manns et al. (1979) found that for each experimental subject there was an optimum physiologic muscular elongation, where the masseter muscle develops the strongest force with minimum EMG activity (Fig. 46).

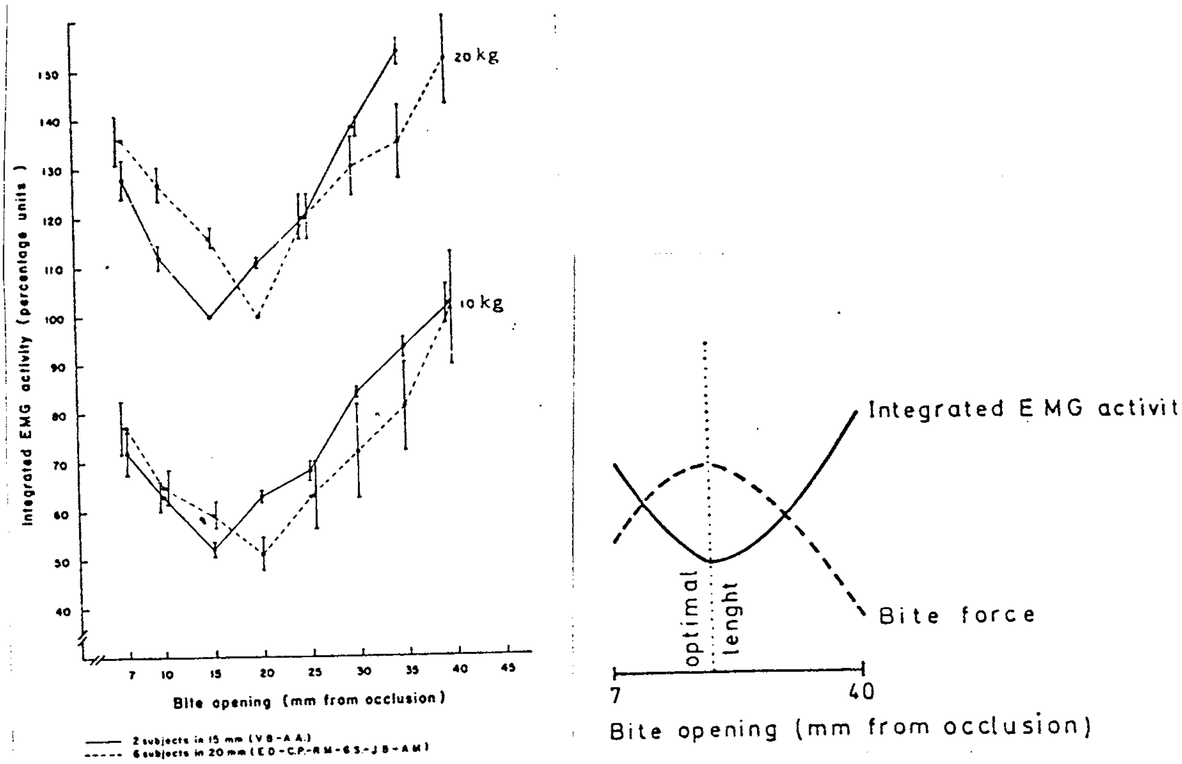


Figure 46 Relation between bite force and integrated EMG activity (Manns et al. 1979)

The linear relationship between biting forces during isometric contractions and integrated EMG has been demonstrated by Garret et al. (1964), Ahlgren (1966) and Ahlgren and Owall (1970),

Pruim et al. (1980) combined several approaches in a study using human subjects clenching symmetrically on a bar placed transversely across the jaws at the levels of the first premolar, the first or second molar. They measured the EMG activity in selected muscles for different bite forces. After making several assumptions they calculated from the equations for static equilibrium , the magnitude of the bite force and the force each muscle was producing. They reported that maximum bite forces are most often exerted in the first molar region, where maximum contraction of all muscles leads to a force resultant in the direction required for static equilibrium. They have also reported that the range of maximum bite force applied to the first molar was 609N to 1308N (both sides summed), second molar 403N to 1206N, whereas the maximum force applied to the central incisors was 386N to 908N which was higher than the previously reported values.

7.4 Barbanel Analysis:

Barbanel's work (1972) was the first analysis of the mandible which contained the basics of the modern biomechanical theories. Even that analysis was restricted to two dimensions and the predictions made were incomplete, important insights were gained for future mathematical/biomechanical models.

Barbanel (1972) used a minimization principle to predict which muscles are utilized to produce a specified bite force. In two different models (1972 and 1981) he required that either the total joint reaction or the sum of the tensions in the muscles should be a minimum (Fig. 47 and 48).

The mandibular muscles considered in the analysis were temporal, masseter, medial and lateral pterygoid muscles. The lines of action of these muscles were determined by dissection. In order to include the influence of deep fibres of the muscle in the analysis a dissection was undertaken to determine the position of centroids of area of origin and insertion. The assumption was made that the line joining these points was the line of action of the muscle.

The results obtained from the alignment of the elevator muscles were compared with those of Mainland and Hiltz (1934),(TABLE IV).

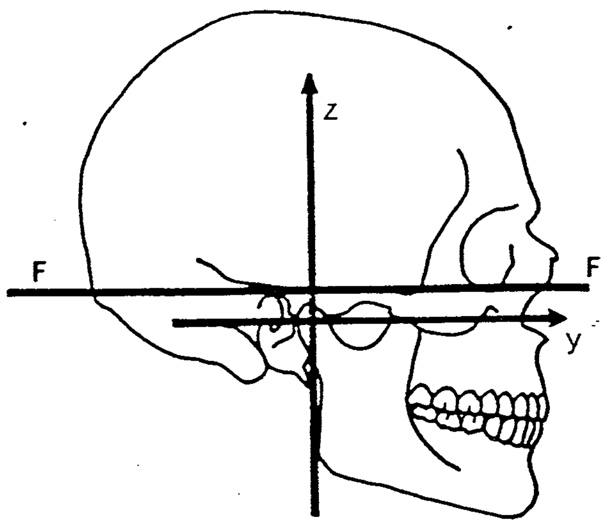


Figure 47 Co-ordinate axis used in joint force analysis of Barbanel (1972,1981), the x axis is directed inwards along intercondylar axis. The y axis is parallel to the Frankfort plane (F-F).

TABLE IV

Muscle force parameters according to Barbanel (1972).

	x	y	z	moment arm (a) cm
MASSETER	-0.12	0.55∓0.04	0.81∓0.02	2.7∓0.2
TEMPORALIS	-0.18	-0.46∓0.03	0.84∓0.01	2.6∓0.17
MEDIAL PTERYGOID	0.38	0.5∓0.05	0.78∓0.04	2.3∓0.17
LATERAL PTERYGOID	0.39	0.91	-0.17	0.0

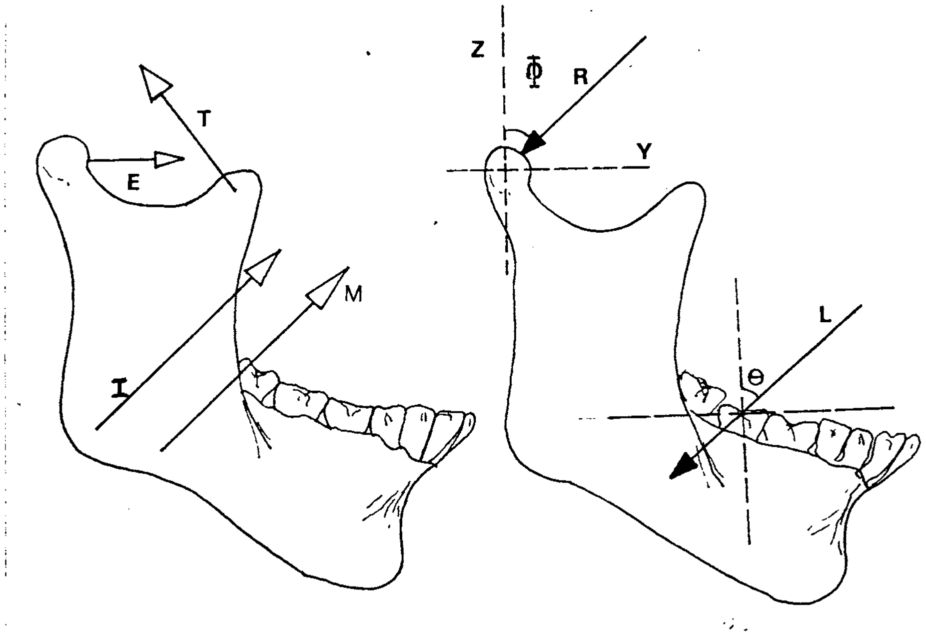


Figure 48 Muscles considered in the analysis (a), occlusal load and joint force parameter (b), (redrawn from Barbanel 1972, 1983).

The forces acting on the mandible at the stage of biting when the jaw is stationary were given as:

The occlusal load: Barbanel (1983) assumed that this load has a magnitude of L, and directed at an angle Θ to the z axis. The L load has a moment about the x intercondylar axis of:

$$L\,(y.\cos\,\Theta + z.\,\text{Sin}\,\Theta)$$

y and z are the co-ordinates of the point ,where the line of action of the occlusal load L intersected the occlusal plane.

The temporomandibular joint force resultant was assumed to have a magnitude R and to act at an angle Φ to the z axis. All the forces were assumed to be symmetrical with respect to the mid-line and to have an equal magnitude on the right and left side of the mandible.

When the mandible is at rest under the action of these forces, the three equations of equilibrium can be written:

Fz components:

$$0.84\,F_M + 0.84\,F_T + 0.78\,F_I - 0.17\,F_E - R\cos\,\Phi - L\,\cos\,\Theta = 0 \qquad 7.1$$

Fy components:

$$0.55\,F_M - 0.46\,F_T + 0.50\,F_I + 0.91\,F_E - R\sin\,\Phi - L\sin\,\Theta = 0 \qquad 7.2$$

The sum of total moment:

$$2.7\,F_M + 2.6\,F_T + 2.3\,F_I - L\,(y\cos\,\Theta + z\sin\,\Theta) = 0 \qquad 7.3$$

There are six unknown quantities in these three equations, and the values of these unknown quantities cannot be obtained by the direct solution of the equations.

Linear programming method (Gass 1975) was used to solve this problem. Using this technique, a function which is known as the objective function, may be minimized subject to additional linear constraints.

The objective function chosen was the unweighted sum of the muscle forces. The equations of equilibrium served as the additional linear constraints which required to be satisfied by the solution obtained for the objective function.

The solutions to the equations were explored using an iterative method, in which values were assigned to the parameters associated with the occlusal load, and to ($\cos \Phi$), and the minimum value of the objective function located.

The values of the parameters were then systematically altered to cover the range which was considered anatomically and physically possible and the procedure repeated to locate the minimum value of the set of solutions to the objective function. All the solutions obtained showed that when the sum of the muscle forces was a minimum, the only active muscles were the masseters.

After utilizing minimum muscle force hypothesis, he rejected the solutions because they couldn't satisfy the necessary conditions - that there be an agreement between the muscles predicted as being active and those shown to be active. He further concluded that there was no evidence that either of the minimum principles employed is applicable to physiological function.

7.5 Pruim, Ten Bosch and DeJough Analysis:

In 1978 Pruim et al. described a method to relate the muscles of mastication EMG activity to static bite forces. Bite forces were measured bilaterally in several positions on the human dentition by means of small wedge shaped transducers. Electromyographic methods were used to derive a relative measure of the activity in the depressor and elevator muscles. They reported that there was no doubt a linear relationship between IEMG activity and the force exerted by individual muscles in isometric conditions did exist. Their main conclusion was that alinearity of individual muscle EMG activity with respect to the external force is or may be due to increased action of antagonists at high bite forces and the role of the opener muscles as antagonists should not be neglected in a muscle force analysis.

In 1980 Pruim et al. chose a mechanical approach to calculate all muscle forces and the forces in the temporomandibular joints during bilateral biting at three different locations on the human dentition. Bite forces were measured bilaterally by means of two calibrated steel transducers. EMG activity was measured by using surface electrodes (activity in the lateral pterygoid muscles were not recorded).

For geometrical data standardized lateral and frontal x-ray cephalograms were used. During the bite force measurement, condyle was located opposite the tubercle and not in the articular fossa (Fig. 49).

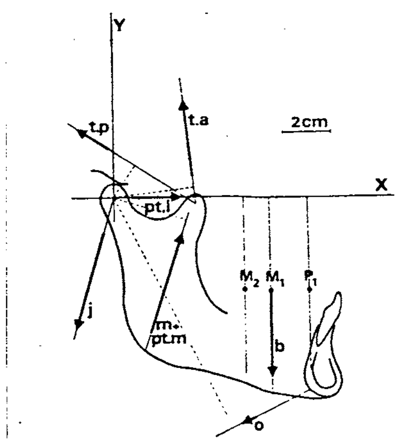


Figure 49 Mathematical model based on Pruim et al.(1981)' s work. The origin was chosen to coincide with the condylar centre, and x-axis is parallel to the occlusal bite plane.

Several geometrical and anatomical assumptions were made for the mathematical model. Forces exerted by contracting muscles were represented by vectors. The direction of these vectors were defined by the connecting lines between the centroids of the origins and the insertions of the muscles. Descriptions of these areas of attachment were derived from the anatomical literature (Fick, 1904; Schumacher, 1961; Sicher and DuBrul, 1975). Forces of the masseter and medial pterygoid muscles were represented by a single vector. Most importantly to minimize the vertical component, lateral pterygoid was assumed to act in x direction.

With respect to electrical activity and to muscle force they have assumed that masseter EMG activity was also representative for the medial pterygoid muscle. The activity in the floor of the mouth was representative for the mouth openers.

The maximum force exerted by a muscle during an entire experimental session was assumed to be equal to the physiological maximum force of that muscle (F_m (max.)).

For each experimental subject a muscle independent value Γ in Pascals was postulated. This value was related to the maximum force exerted by a muscle to its physiological cross section Φ_m in m^2.

$$F_m(\text{max}) = \Phi_m \cdot \Gamma \qquad 7.4$$

The values of the physiological cross sections were as determined by Schumacher (1961) for males according to Buchner's (1877) method, corresponded to the values for the test subjects (Appendix III). The muscle force exerted during isometric contraction F_m was directly proportional to the integrated EMG (Lippold, 1952; Bigland and Lippold, 1954; Bergstrom, 1959; Goubel, 1971; Cnockaert et. al., 1975; Yemm, 1977; Pruim et al. 1978).

The relationship between the EMGI and muscle force was expressed as:

$$F_m = \frac{EMGI_m(\text{actual})}{EMGI_m(\text{max})} \cdot F_m(\text{max})$$

$$F_m = \frac{EMGI_m(\text{actual})}{EMGI_m(\text{max})} \cdot \Gamma \times \Phi_m \qquad 7.5$$

$EMGI_m$(max) was the maximum value of $EMGI_m$ of a single muscle ever recorded from an electrode pair during a complete experimental session. The maximum EMG - activity in the floor of the mouth was recorded, prior to their test procedure.

For muscle physiological cross sectional areas Schumacher's (1961) values were used. For the temporal muscle the value was divided into posterior and anterior portions according to a ratio derived from Carlsoo (1952),(Appendix III).

Since there was no quantitative data available for the antagonistic muscles, four pairs of human anterior digastric muscles, were measured according to Schumacher's (1961) and Honee's (1970) methods. The results were variable and they have reported a mean cross section value per side of 0.8 cm^2, standard deviation of 0.2 cm^2. A physiological cross section of the antagonistic muscles of 1 cm^2 was assumed to be a close estimate for the total openers and was used in their calculations.

They have combined the corresponding forces acting on left and right halves of the mandible and projected the force vectors on a sagittal plane.

The laws of static equilibrium were used in calculating the force components and moment equilibrium equations.

$$\Sigma\, F_m \cdot a_m + F_b \cdot a_b = O \qquad 7.6$$

where F_m is the muscle force exerted during isometric contraction, F_b bite force, a_m moment arm of the muscle force, and a_b moment arm of the bite force, according to the condyle centre.

When the moments are referred to the origin and equation (7.5) is used, this leads to:

$$\Sigma\, \frac{EMGI_m(\text{actual})}{EMGI_m(\text{max})} \cdot \Gamma . \Phi_m \cdot a_m + F_b \cdot a_b = O \qquad 7.7$$

$EMGI_m$ values and the sum of both bite forces were measured in each bite. Using the geometric and physiological data, muscle independent value was calculated using the equation (7.7). After substitution of Γ in equation (7.5) all forces in the levator and antagonistic muscles during that bite were calculated.

Joint forces (F_j) and lateral pterygoid muscle forces (F_p) were calculated using the equations (7.8) and (7.9).

$$\Sigma F_m . \sin \alpha_m + F_j . \sin \alpha_j - F_b = O \qquad 7.8$$

$$\Sigma F_m . \cos \alpha_m + F_j . \cos \alpha_j + F_p = O \qquad 7.9$$

Averaged maximum muscle forces calculated with this method, when reduced to one side are higher than the data on theoretically calculated forces as given by Schumacher (1961) for all levator muscles and those for lateral pterygoid muscles as given by Honee (1970). All muscle forces as given by Carlsoo (1952, 1956) were higher than Pruim's (1980) values (Appendix III).

The values of maximum muscle tension Γ, found in Pruim's investigation, were much higher than the often used value of 1×10^6 N/m^2, originating from Johnson (1903) as cited by Fick (1910), (Appendix III).

Comparative data on maximum bite forces showed that the highest bite forces and muscle forces were most often exerted in the first molar region, where according to investigators, maximum contraction of all muscles leads to a force resultant in the direction required for static equilibrium.

Even that the method of analysis was only two-dimensional, application of this mechanical model to a group of subjects reconfirmed the validity of the method to study force patterns.

CHAPTER VIII

FINITE ELEMENT ANALYSIS IN BIOMECHANICS

When a structure is loaded, stresses are generated in its materials. The distribution of these stresses, their magnitudes and orientations throughout the structure, depend not only on the loading configuration, but also on the geometry of the structure and the properties of its materials. In addition, the stresses are influenced by the interaction of the structure with its environment and by physical conditions at boundaries between different materials. In a theoretical stress analysis the stress distribution is evaluated by using a mathematical model. In such a model, that represents the real structure to a certain degree of refinement, the structural aspects such as loading, geometry, material properties, boundary and interface conditions are described mathematically. The mathematical descriptions can be more or less accurate, depending on the level of refinement required, and are usually based on experimentally determined data. In the solution procedure, the structural descriptions are combined in mathematical equations based on theories of solid mechanics and these are solved to give the stresses.

Various solution methods are available in classic mechanics for certain classes of structures. The Finite Element Method, as a theoretical method of structural stress analysis, however, is suitable in principle for any structure. Geometries, material properties and loading of arbitrary complexity can be accounted for.

In using the method, the model as a geometric entity is defined first. This model is then mathematically divided into a number of blocks (elements), connected at specific locations, called nodal points or nodes. The boundary conditions and loading configurations are numerically defined, in boundary nodes. Every element is assigned one or more parameters that define its material behaviour.

The material properties of the actual structure can be isotropic or anisotropic and can vary from element to element. A uniformly distributed load over the surface of an element needs to be reallocated to the corresponding nodes of that element as the technique provides a load-displacement relationship at the nodes only, Zienkiewicz (1971).

The key to the finite element solution is in evaluating the stiffness matrices of all elements individually by choosing a proper interpolation function. It defines displacement of a point within an element in terms of the nodal displacements. The proper interpolation function, also known as shape function, ensures compatibility on the boundary surface between elements and convergence of the solution. The displacement of a point within the element is given as

$$u(x,y) = [N]\,[u]^e \qquad (8.1)$$

where N is a matrix of shape functions, while $[u]^e$ is the displacement of the nodes of an element.

Further the following relations are applicable to small strains

$$[\epsilon] = [B]\,[u] \qquad (8.2)$$

$$[\sigma] = [D]\,[\epsilon]$$

$$[\sigma] = [D]\,[B]\,[u] \qquad (8.3)$$

Where matrix [B] is the partial derivative of the matrix [N] and matrix [D] is the elasticity matrix for the material of the element.

The principal of virtual work, the minimisation of energy or the weighted residual technique may be used to obtain the load displacement relationship for an element.

This is given by:

$$[P]^e = \int_V [B]^T\,[D]\,[B]\,.\,dv\,.\,[u]^e$$

$$= [k]^e\,.\,[u]^e \qquad (8.4)$$

where;

$$[k]^e = \int_V [B]^T\,[D]\,[B]\,.\,dv$$

where $[B]^T$ is the transpose of matrix [B], and $[k]^e$ is the stiffness matrix for an element.

The stiffness matrices $[k]^e$ of the elements are assembled to form an overall stiffness matrix, [K]. This is related to the overall load vector, [P] and displacement, [U] by:

$$[P] = [K]\,[U] \qquad (8.5)$$

This set of simultaneous linear equations is solved to determine the unknown displacement field and thereafter stresses are evaluated using equation (8.3).

The procedures described in equations 8.1 to 8.4 are carried out automatically in finite element programs. The program calculates the stiffness characteristics of each element and assembles the element mesh through mutual forces and displacements in each node. The computer time needed depends progressively on the number of elements applied, and on the element type (Fig.50).

In interpreting the FEM results we must differentiate between the validity of the model, (the precision by which the entity of methematical descriptions of structural aspects such as loading, geometry, material properties, boundary and interface conditions represents the real structure), and the accuracy of the model, (the precision by which the FEM mesh can approximate the exact solution for the model). The accuracy of the model can be tested by increasing the number of elements or benchmark tests on similar structures for which the exact solution is known. The validity of the model must be assessed by experimental verification or other means.

A variety of element types are usually available for 3-D and 2-D structures in an FEM computer package, different in their number of nodal points and their shapes. The computer time required for 3-D elements is many times higher as compared to 2-D elements.

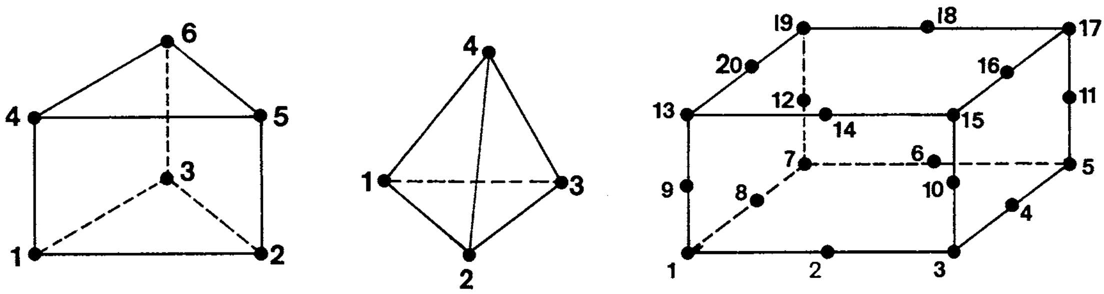


Figure 50 Three dimensional solid elements.

FEM programs are usually applied in combination with pre processors and post processors, computer programs that handle the element division and the graphical representation of the results, respectively.

The finite element method, after about ten years from its initiation (Turner, 1956) in stress analysis of structures in engineering mechanics was first used by Wiederhielm et al. (1968) for the structural response of arterioles and later by Kobayashi et al. (1971) for the stress analysis of the corneo-scleral shell, Brekelmans et al. (1972) for the mechanical behaviour of skeletal parts and Tesk and Widera (1973) for the stress distributions in oral implantology.

Finite element analysis is presently being used to investigate the stress related architecture of bone and bone remodelling processes, to test and to optimize artificial joint designs and fracture fixation devices, and to study the mechanical behaviour of tissues such as articular cartilage and intervertebral discs.

In its initial years of introduction to the biomechanics field, the FEM has sometimes regarded as a magic tool to solve all problems. Few scientists in the field were specialised in its use or aware of its capabilities and limitations. Suspicion became evident when a number of optimistic users appeared to rely on it as a tool to generate solutions without carefully formulating the problem. The complexity of biological structures was often underestimated. In particular, little was known about joint and muscle loading, and the rheologic properties of bone and connective tissues, let alone their stress failure mechanisms and biologic reactions to stress.

The practical potential of the FEM for modelling complex structures adequately was often overestimated. Suited in principle for structures of arbitrary complexity, its application to three dimensional and/or nonlinear structures is still tedious and expensive. Although these limitations are gradually diminishing through biomechanics research, computer software and advances, the time and efforts this would require were not fully appreciated during the early stages.

In all fairness, however, it must be said that the greater part of the efforts in this second decade of the FEM in biomechanics were method oriented rather than problem-oriented. Progress was made towards closer and more precise definitions of problems, and a more balanced multidisciplinary approach to their solutions. Although many analyses have not led to solutions of realistic problems directly, they have served indirectly in demonstrating advances in methodology to the biomechanics field and in guiding experimental biomechanics research , for instance, in the areas of materials testing and functional force evaluation (Huiskes, 1983).

FEM analysis are based on theories of continuum mechanics. The continuum materials involved in the analysis of bones are cortical (or compact) bone, and trabecular (or spongeous) bone. No material is continuous at any level, but trabecular bone is even noncontinuous on a macroscopic level, hence it is a structure rather than only a material.

In quasi-static loading both cortical and trabecular bone, although anisotropic and nonhomogeneous, behave elastilyc by approximation. Their elastic properties depend on such factors as age, species and mineral content; hence, variability between subjects can be high. It must be appreciated that bone is a biological tissue. Its structure at any moment represents an equilibrium between continuous processes of resorption and formation, which may be affected by outside stimuli of a chemical, mechanical and electrical nature. There are many indications that bone architecture is influenced by stresses or strains, a relation expressed in "Wolff's Law" (Wolff, 1870; Koch, 1917), which is actually a hypothesis (Hayes et al., 1982). Excellent reviews on the biomechanics of bone were edited by Cowin (1981). Bone loading due to muscle and gravitational forces is dynamic and little is known regarding its real characteristics and magnitudes (Chao and An 1982).

The femur is the bone most frequently analyzed due to historical developments and its common involvement in orthopaedic treatments, such as prosthetic replacement of the hip joint. The early FEM was adopted for stress analyses of the femur by Brekelmans et al. (1972), and Rybicki et al. (1972). As most of the early efforts, their analysis were not so much directed at a specific problem, but aimed at demonstrating the possibilities of the FEM. Both applied 2-D plane stress elements of uniform thickness, although Rybicki et al., analyzing the proximal part of the bone only, accounted for non-uniform thickness by varying the Young's moduli of the elements. Results were compared with those of a 2-D beam model analysis, yielding good agreement in the diaphyseal part only. A comparable model was used by Wood et al. (1973), who applied 2-D elements by variable thickness.

The earlier investigations of the mandibular system by Tesk et al.(1973) utilized two-dimensional plane stress, plane strain linear elastic finite-element analysis. In these studies, all bone was treated as being isotropic and homogeneous, with the boundaries of the model of the implant tissue system extending essentially to infinity in the lateral direction.

8.1 Finite Element Analysis in Orthopaedic Systems:

Early 3-D FEM models of the femur were those of Scholten (1975) and Olofsson (1976), followed by Valliappan et al. (1977), Harris et al. (1978) and Rohlmann et al. (1982). The problems associated with data manipulation, interpretation and representation, frequently more demanding in 3-D FEM analysis, were clearly demonstrated in these publications. Some models contained a high degree of sophistication (Fig. 51). These earlier investigations were, in fact, all method oriented, focusing on the application of the FEM itself, with little emphasis on problem-oriented solutions within a specific investigation. All in some way compared results of different approaches such as beam, 2-D, 3-D, or experimental analysis techniques. The data generated for the shaft of the femur compared favourably among different mathematical models. Deviations were more extensive, however, in the proximal region.

The early 3-D model by Scholten (1975) is a refined one, even according to present standards, with probably the highest mesh density to date. The models presented by Valliappan et al. (1977) are rather rough in comparison, but consist of higher order elements. Only these authors report an extensive convergency test by increasing the number of elements to check the adequacy of the mesh.

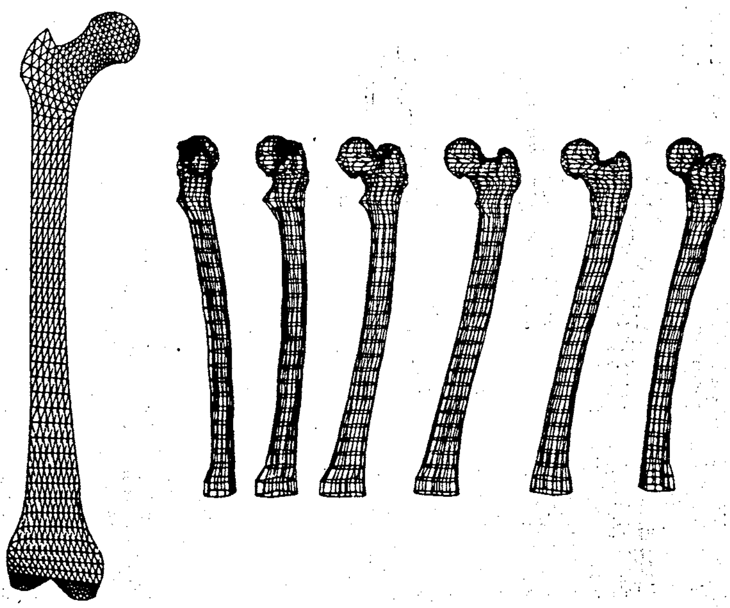


Figure 51 Element meshes of the human femur; 2-D mesh of 936 triangular 3-node elements (537 nodes)(Brekelmans et al. 1972); 3-D mesh of 1950 8-node isoparametric hexahedron elements (2532 nodes)(Rohlmann et al.,1980).

Few authors have correlated theoretical and experimental results in detail. Valliappan et al. (1977) compared FEM results with those of stress-coat experiments, revealing good agreement in a relative sense but poor in an absolute sense. The same conclusions was reached by Rohlmann et al. (1982), based on a comparison between 3-D FEM and strain-gauge experimental results from paired femoral bones.

The mechanics of the femoral head in the presence of avascular necrosis - a local degeneration of the trabecular bone - were studied by Brown et al. (1980). This was a good example of a more problem-oriented approach in which the FEM can play a significant role. Other examples of problem-oriented studies, employing relatively simple FEM models, are those pertaining to bone growth and remodelling. Hayes et al. (1978, 1982), have attempted to correlate the trabecular bone architecture of the human patella and its stress distribution for the quantification of Wolff's hypothesis. Using a 2-D FEM model, they found that high Von Mises effective stress correlates with regions of high trabecular density, and they re-established that the trabecular structure aligns itself with principal stress orientations. They also demonstrated that anisotropy of trabecular bone hardly affects the stress distribution if its nonhomogeneity is correctly accounted for.

A number of FEM stress analysis of bones have been designed for problems of fracture, fracture fixation, and bone remodelling.

Two dimensional FEM models considering the femoral hip prosthetic component were analyzed by Hampton et al. (1976, 1981), Andriacchi et al. (1976), Svensson et al. (1977), Kwak et al. (1979), Yettram and Wright (1979, 1980), Cook et al. (1980), Sih et al. (1981), and Skinner et al. (1982). Varying degrees of mesh refinement were used, different phosthetic designs simulated, and several questions addressed. Influences of varus-valgus placement, acrylic cement layer thickness, stem shape and material were investigated principally. In the majority of these FEM analyses, linear elasticity, isotropy and homogeneity of cortical and trabecular bone were assumed, whereas interfaces were modelled as rigidly bonded. Realistic nonhomogeneity of trabecular bone, based on experimental data, was taken into account by Roehrle et al. (1977) and Scholten et al. (1978). The effects of anisotropy in cortical bone were investigated by Valliappan et al. (1980) and Vichnin and Batterman (1982,1986).

The nonlinear effects of cement-metal interface loosening were studied by Svensson et al. (1977), Huiskes and Schouten (1980a) and Hampton et al. (1981). It became evident from these 2-D analysis that slip and tensile loosening at the element connection has a highly significant effect on cement and interface stresses, whereas those in the stem and the bone are less seriously affected.

An informative study was performed by Rohlmann et al. (1980) to evaluate the accuracy of a 3-D FEM model, as compared to experimental strain-gauge measurements on a laboratory model, related to a femoral THR structure. A prosthesis was fixed with acrylic cement in a metal tube. This experimental model was investigated with the 3-D FEM, using a mesh of moderate refinement (8 node- 232 hexahedron elements), (Fig.52) when compared to other analysis published in this area.

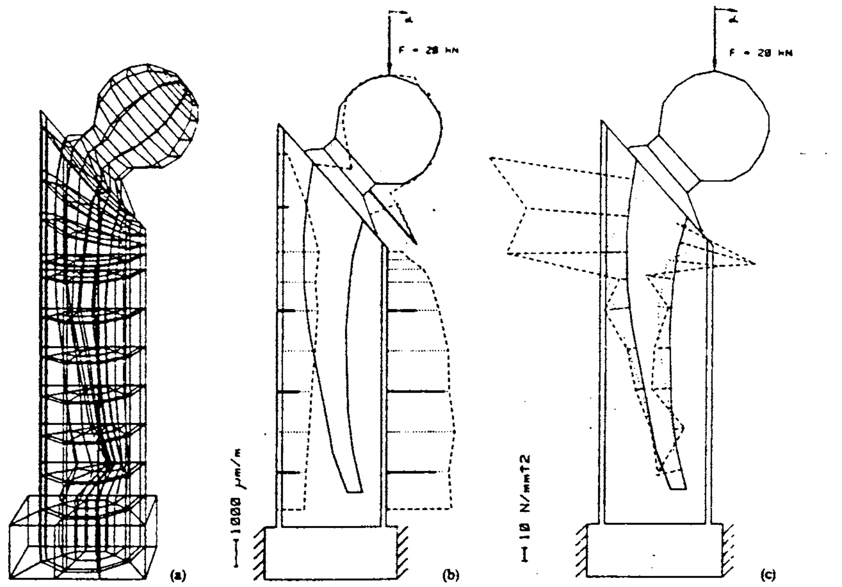


Figure 52 A 3-D FEM analysis of a physical model ,simulating a femoral THR fixation structure (Rohlmann et al. 1980).

Although the trends established were more or less equal in both cases where the most significant stresses on the "medial" side of the tube were concerned, they found considerable differences between the experimental and theoretical stress results at certain locations, specifically the "lateral" side of the tube which simulates the bone. They attribute these deviations mainly to interface loosening, occurring in the experimental mode and not simulated in the FEM model.

An improved model was carried out by Svensson et al. (1980). Their experimental configuration was in principle the same as described above, but strain measurements were taken on the prosthesis in this case. Their mesh density was somewhat less (167 elements), but their elements were more sophisticated (20 node isoparametric hexahedrons). They compared experimental stresses in the stem with those of theoretical results, assuming bonded surfaces and later allowing slip to occur at the cement-stem interface. In the first case agreement was poor (up to 40% lower in the FEM results), in the second case it was remarkably good.

In an investigation to determine the effects of the presence of two lengths of proximal Muller prosthesis on predicted failure loads as compared to those for an intact femur was carried out by Vichnin and Batterman (1986). Three-dimensional stresses in a bone-cement-prosthesis system were determined, with both isotropic and transversely isotropic material properties used for the diaphyseal cortex. Significant increases in prosthesis stem stresses were found when the transversely isotropic material properties were employed in the diaphyseal cortex. They have further concluded that accurate anisotropic material properties for bone are essential for precise stress determination and optimum design in prosthetic implants.

Anand et al. (1976) and Ducheyne et al. (1978) studied the influences of bone ingrowth in porous watings of intramedullary fixated stems, using axisymmetric FEM models in which the elastic moduli of the coatings were adapted to account for the ingrown material.

Seidelmann et al. (1982) investigated the head-stem connection in prostheses with ceramic heads, using 2-D and axisymmetric models.

Stress analysis of the acetabular cup component in total hip replacement have become popular only recently, probably due to the relatively late occurrence of cup loosening and the complicated geometry of the pelvis. A 2-D FEM model of this structure was developed by Vasu et al. (1982) to demonstrate changes in the acetabular stress pattterns after joint replacement. Carter et al. (1982), using the similar models and Bartel et al. (1985), studied effects of adding a metal encasement over the plastic prosthetic component and also evaluated the influences of acrylic cement thickness. Three dimensional FEM models as reported by Oonishi (1981), Goel et al. (1981), and Oonishi et al. (1983) have better potential in modelling the complicated acetabular geometry in principle.

A relatively new hip joint replacement, known as the "resurfacing" prosthesis, contains a metal cup placed over the femoral head to replace its articular surface. The FEM was used to evaluate this procedure in a 2-D model by Shybut et al. (1980) and in an axisymmetric model 3-D loading and stress fields by Huiskes and Heck (1981).

The axisymmetric model (Fig.53) better represents the 3-D structural integrity of the metal cup, however, realistic inhomogeneity of trabecular bone can only be included in the 3-D model.

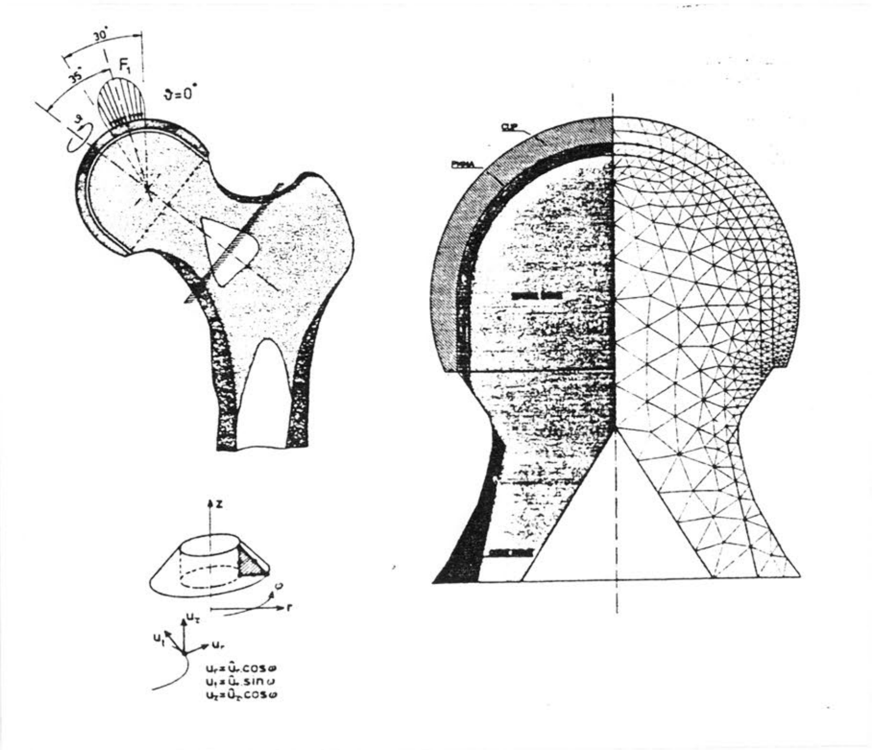


Figure 53 FEM analysis of the femoral component of a hip surface replacement.Note the local mesh refinement in regions where high stress gradients are expected(Huiskes and Heck, 1981).

FEM analysis in artificial knee joint design and fixation become more frequent in the late seventies. With some exceptions, all studies have dealt with the tibial component fixation exclusively, which indeed is the part giving the most problems in patients. The tibial and femoral stem fixation of hinged prostheses were analyzed in 3-D models by Roehrle et al. (1980, 1982), and the tibial fixation alone in axisymmetric models, taking 3-D loading into account by Campen et al. (1979), Croon et al. (1982). The tibial plateau fixation of non-hinged surface replacement prostheses was investigated using 2-D models by Lewis (1977), Hayes et al. (1978), Vichnin et al. (1979), and Askew and Lewis (1981).

Most of these studies were directed at the function of the central post, used to fixate the plateau in the tibia (Fig.54).

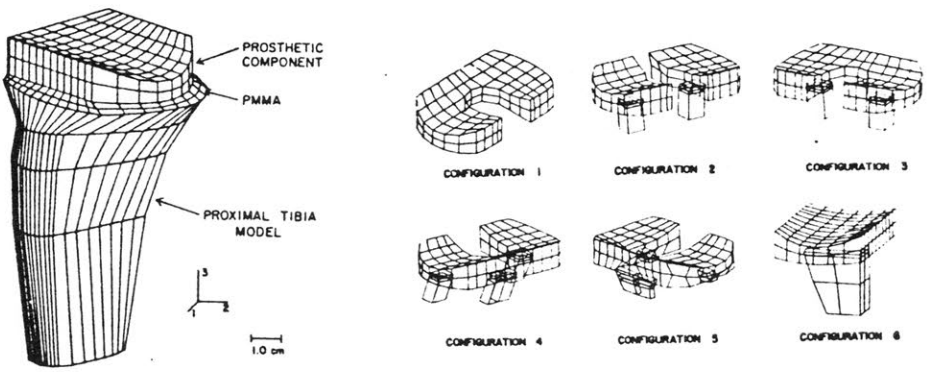


Figure 54 A 3-D FEM model of 768 eight node linear isoparametric hexahedron elements to investigate the tibial plateau fixation of total knee replacement (Lewis et al.,1982).

The Lewis et al. (1982) model specifically is quite anatomical, including nonhomogeneous trabecular bone properties.

An axisymmetric FEM model of the tibial plateau fixation without central post was reported by Shrivastava et al. (1980), whereas axisymmetric geometry taking 3-D loading into account was studied by Murase et al. (1983). Hori et al. (1982) have investigated the effects of a soft tissue layer between the tibial implant and the bone on the local stress distribution and fixation strength, using a nonlinear FEM approach. Such a soft tissue line can often be seen to appear in postoperative roentogenograms after sometime. It was shown to have a significant influence on interface shear stresses in their study.

In revising failed total joint replacements with extensive bone loss and for bone and joint reconstruction after tumour resection, the use of "custom fit" prostheses is increasingly in demand. In the design and selection of these devices, the FEM has been used by Chao (1980) in an axisymmetric model to evaluate a specific type of ceramic prosthesis based on extra cortical fixation, which has been applied in shoulder prostheses.

Analysis of prosthetic components other than in the hip and knee joints have been rare. Joints that are also replaced relatively frequently are those in the hand. An axisymmetric FEM analysis (3-D loading) of a intramedullary fixation system for finger-joint prostheses was carried out by Huiskes et al. (1980). Stress results were compared in detail with histological findings in animal experiments in a related study, and some corresponding effects were found.

8.2 Finite Element Analysis in Oral Systems:

The application of the finite element stress analysis technique, to the biomechanical investigations of the oral systems such as human tooth, periodontal membrane, implant design and mandibular bone modelling, surprisingly has not been initiated until the early 1970's.

Starting with the early experimental work by Burstone (1962), Davidian (1971) developed and used a first-order computer model to examine the distribution of stresses along the root of a maxillary central incisor after an external force was applied to the crown. Using several simplifying assumptions regarding the periodontal membrane, Davidian calculated the stress and displacement at several levels in the periodontal membrane and the centre of rotation of the tooth. The theoretical centres of tooth rotation compared well with those measured by Burstone.

This work was extended by Daly et al. (1972), who developed a finite element mathematical model of the human central incisor-periodontal ligament (PDL) structure and analyzed this model using numerical computer analysis procedures. The results of their work, supported by clinical experiments, have led to a better structural characterization of the PDL and an improved understanding of tooth mobility.

In 1973 Thresher and Saito using two dimensional plane strain finite element method, modelled the maxillary central incisor (Fig. 55). They used the simple triangular elements and applied a concentrated force to the tip of the tooth in the transverse direction subjecting the tooth to a bending stress. They have reported that, the major portion of the load was carried by the enamel portion of the tooth. They concluded that a three dimensional analysis was necessary for teeth of more complicated geometry. Farah et al. (1973) used both the photoelastic and the finite elements methods to study the stress distribution in the first molar with a full crown preparation. They adopted an axisymmetric idealization and used the triangular elements for discretization. In their model the tooth was loaded axisymmetrically with a 50 kg load. They have concluded that the finite element axisymmetric model yields an approximate yet reliable information concerning the stress distribution in the first molar compared to the photo elastic method.

In order to illustrate the necessity of using a three dimensional analysis for a more realistic stress distribution in the first molar, a two-dimensional axisymmetric and three dimensional analysis were carried out by Goel and Valliappan (1976). For the axisymmetric analysis, isoparametric cubic quadrilateral elements whereas both linear and quadratic brick elements were used for the three-dimensional analysis.

While the loading pattern in the first case was axisymmetric, the same loading over a similar area was applied in the three dimensional analysis for the purpose of comparison. An arbitrary load of 25 kg was assumed to be acting vertically.

They have reported that at all sections of the models the maximum stresses given by the axisymmetric model are much higher - approximately 30 percent - than those predicted by three dimensional analysis. They have further concluded that the difference in the stress distribution predicted by the "quadratic" and "linear" element models in the three-dimensional analysis is negligible indicating that the results obtained by either model are acceptable as long as approximately the same total number of nodes are used in the discretization.

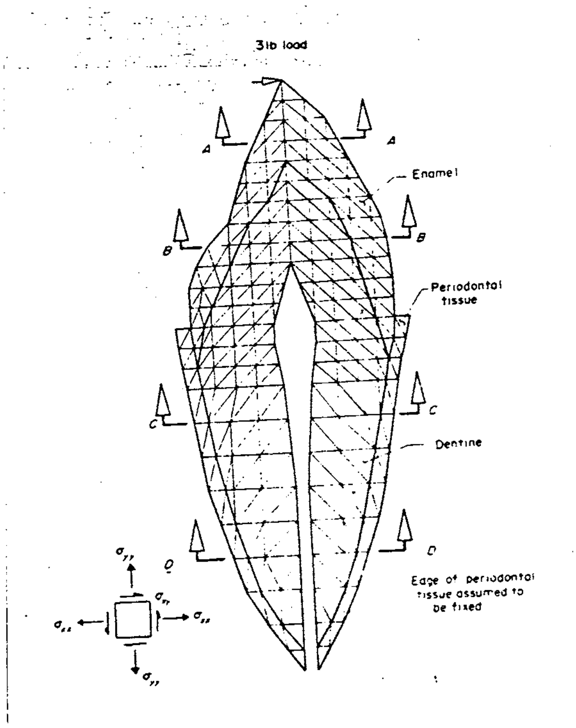


Figure 55 A finite element tooth model (Thresher and Saito, 1973)

While contributions to this expanding base of biomechanics methodology are being made few studies directly relating to oral implantology have also emerged. Tesk and Widera (1973) have analyzed the effects of the states of stress arising from matched and mismatched elastic moduli between bone and dental implants. A two-dimensional finite element analysis was used to investigate the distribution of stress in the alveolar bone surrounding a vitallium blade implant. (Fig.56). Their results showed areas of concentration of high tensile stress in addition to compressive stresses. In their conclusions they stressed the need for studying the biomechanical effects of implant geometry in order to determine if bone resorption may possibly be induced by the design shapes of dental implants.

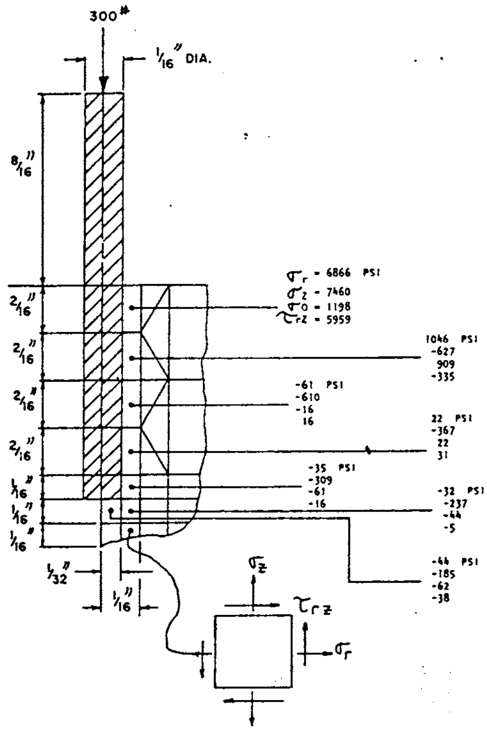


Figure 56 Basic model of implant in bone (Tesk and Widera, 1973).

Using similar models and methods Privitzer et al. (1974), and Widera et al. (1976) utilized both two-dimensional plane stress, plane strain, and three-dimensional axisymmetric linear elastic finite element analysis, (Fig.57). In these studies, all bone was treated as being isotropic and homogeneous, with the boundaries of the model of the implant-tissue system extending essentially to infinity in the lateral direction. On the basis of the results obtained, they have reported that an encapsulating periodontal membrane provides very beneficial and necessary effects. The optimum thickness of membrane was shown to be 0.4 mm to provide an approximately tenfold reduction in stress around a tooth or implant relative to one which did not have a membrane. Beyond this thickness, essentially no reduction in stress was reported.

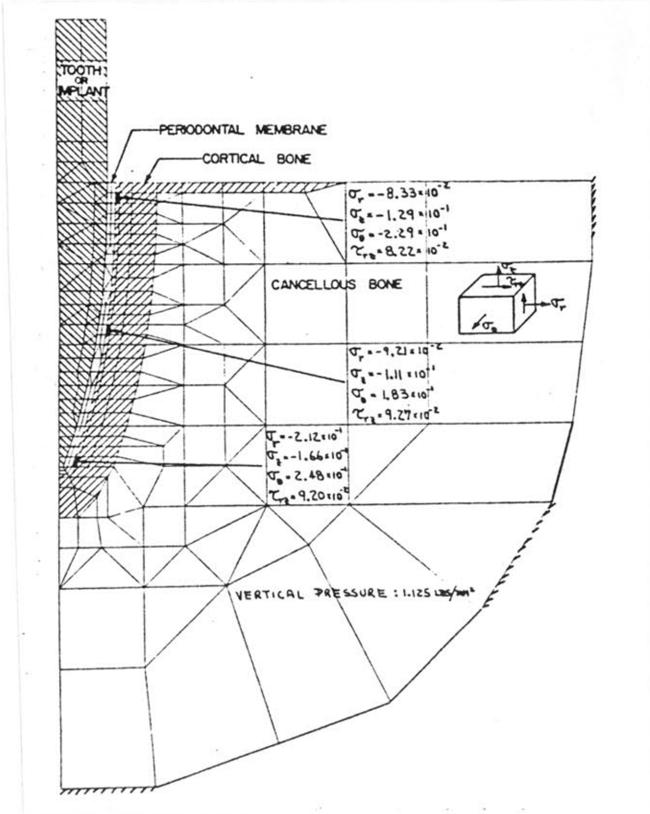


Figure 57 Finite element model of implant, periodontal membrane and the mandible, (Widera et al., 1976).

Based on the anatomical form and dimension, human tooth, periodontal membrane, and mandibular bone was simulated by using two dimensional finite element analysis by Kitoh et al. (1977). They have investigated the effect of the Poisson's ratio on periodontal membrane under the occlusal forces.

In their model, they have assumed that the elastic modulus represented the elastic properties in the periodontal membrane and the Poisson's ratio represented the non-elastic or visco elastic properties. The form and dimension of the model was decided by tracing the sliced human dry mandible at the centre of the lower first premolar. The traced model was divided into 299 triangles with 144 nodal points (Fig.58). Each nodal point was plotted in x-y co-ordinates according to the anatomical dimension of a human enamel, dentin, periodontal membrane, spongy bone, and cortical bone. The mechanical properties of enamel, dentin, spongy bone, and cortical bone were based on the published data. Mechanical properties of human periodontal membrane was investigated under a constant Poisson's ratio of 0.499 and varied elastic moduli. The tooth was loaded 0.04 kg in buccolingual direction.

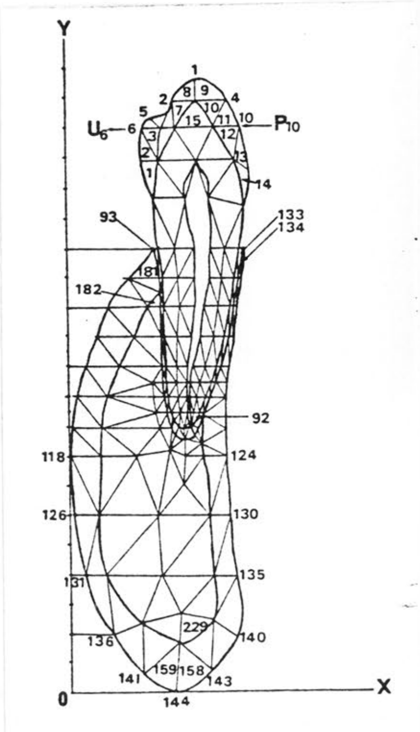


Figure 58 Finite element model of first premolar (Kitoh et al., 1977).

They have reported that the displacement of periodontal membrane is much larger than that of the tooth and mandibular bone under occlusal load. Their analysis did not include the true visco-elastic behaviour of the periodontal membrane, and their results on the displacement in only two dimensional analysis of a single section was not sufficient enough to satisfy the total boundary conditions that exist in the mandibular system.

Despite the fact that these earlier investigations could demonstrate the suitability of finite element method to simulate models equivalent to human joints and for examining the internal stresses and it's displacement, until late 1970's, very little attention has been given to the tooth/implant-periodontal membrane - bone inteface. Implant-bone interface assumptions are critical in the prediction of implant displacement and hence the state of stress surrounding tissue. If the implant-bone biomechanical interaction is not properly characterized design decisions based upon erroneous stress levels in surrounding tissue could be made.

Atmaram et al. (1978) have determined the effect of a pseudo-peridontal membrane at the implant bone interface with finite element analysis. Cylindrical dental implants fabricated from materials with a broad range of elastic modulus were used in the study.

It was found that when a pseudo-periodontal membrane is incorporated in the model, it alone dictates the resulting stress distribution in implant and surrounding tissue, irrespective of the implant material employed, provided that the membrane is less stiff than either the implant or surrounding bone. The results indicated that the incorporation of the pseudo-periodontal membrane had an advantgeous effect on maximum alveolar stress for implants fabricated from vitallium or titanium and a deleterious effect on alveolar stress for vitreous carbon implants.

To determine the implant-bone interface characteristics for bioglass dental implants, three-dimensional finite element analysis was used by Weinstein et al. (1980). The finite element grid system was developed from a geometric analysis of a representative canine mandible; the implant site was modelled to represent that used in the animals. Fourteen buccal-lingual sections which defined the mandibular structure over a length of 25.8 mm or about 10 mm mesial and 10 mm distal from the implant were chosen for construction of the finite element model. A finite element grid system was developed for each of these two dimensional buccal-lingual sections and each of these connected to form the three dimensional network of hexahedral and tetrahedral elements. The model consisted of a total of 461 nodes and 851 constant strain elements. The gridwork was reported to be finest near the implant (Fig.59).

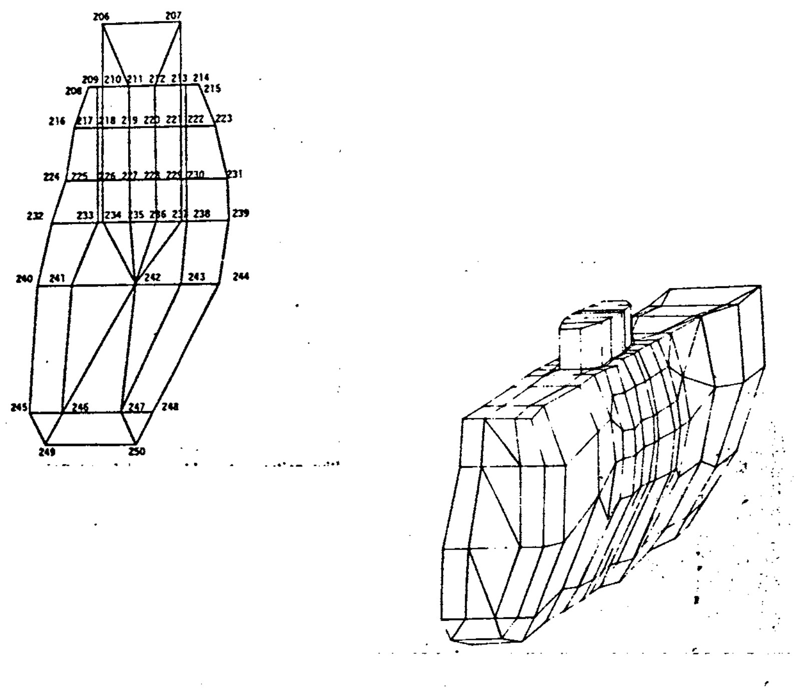


Figure 59 a) Buccal-lingual finite element grid centre selection; b) Three-dimensional finite element model (Weinstein et al., 1980).

The results of their study indicated that the assumption of a discontinous change of elastic properties at the bone-implant interface was poor assumption for the bioglass implants. They further concluded that calculated load-displacement values from the finite element model with soft tissue interface were in excellent agreement with measured results for an implant which exhibited a soft tissue interface region experimentally.

To determine the effects of "trabecular bone representation" on the stress distribution around a dental implant, a two-dimensional finite element model was constructed by Lavernia et al. (1981). The model was constructed to represent the anatomical geometry of a buccal-lingual section through a baboon mandible hosting a blade type dental implant. The microradiographic section which was used to construct the finite element grid and the model is shown in Fig.61 . The finite element grid contained 738 nodal points and 814 constant strain quadrilateral and triangular elements. Each element in the finite element grid was assigned a material elastic property based upon examination of the microradiographic section. All materials were assumed to be linearly elastic and isotropic. Bone was assumed to be in direct contact with the implant surface as determined from analysis of histological sections.

Previous investigators using finite element analysis to model implant systems, have assumed a modulus of elasticity for cancellous bone based upon its apparent density thus not accounting for its true inhomogeneous structure. Lavernia et al. (1981), in their modelling technique, represented a cancellous bone, which was called "a trabecular representation". According to this model, the dark spaces in Fig.60 was represented as intertrabecular space while the white areas was used to represent trabecular bone. In this manner the grid was used to model the inhomogeneous structure of cancellous bone.

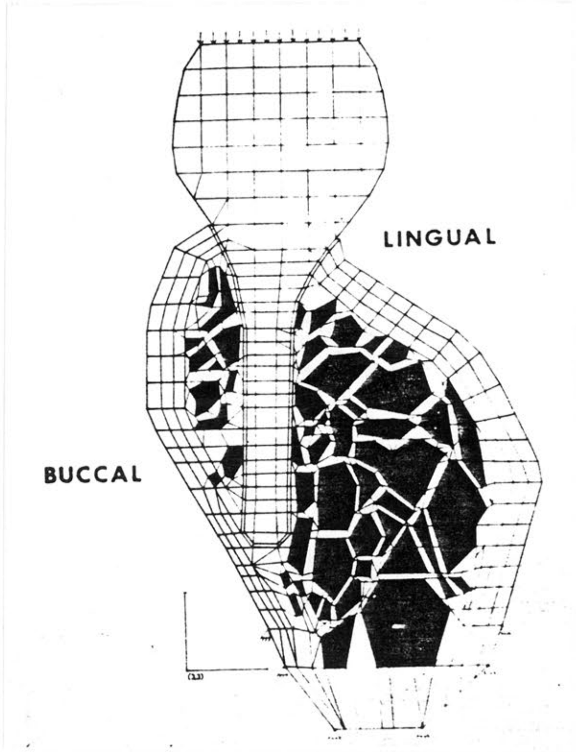


Figure 60 Finite element grid representing a buccal-lingual section through a baboon mandible with dental implant in place (Lavernia et al., 1981).

They have reported that, the type of modelling used to represent cancellous bone-trabecular versus bulk - had a substantial effect on the magnitude of the stresses in the cortical plates; however, the stress profiles for both cases were very similar (Fig.61).

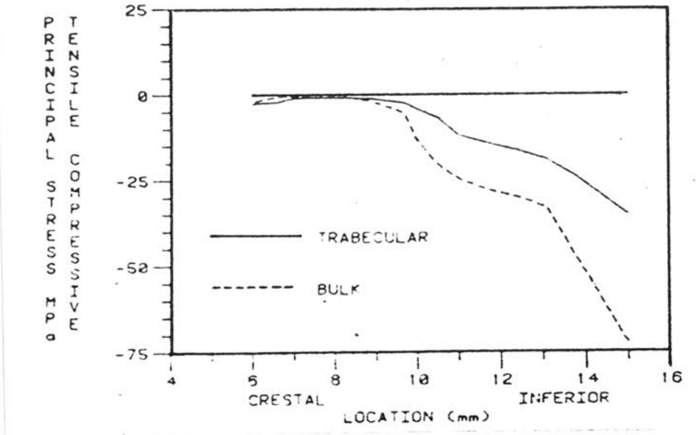


Figure 61 : A crestal to inferior stress profile in the buccal cortical plate contrasting the bulk property and trabecular representation with an LTI carbon implant (Lavernia et al., 1981).

Compressive stresses in the cortical plates reported using the "bulk representation" were generally greater than those obtained with the trabecular representation for both implant materials studied.

To investigate the variations in stress distribution due to the differing infrastructural geometry of a blade type implant, three dimensional finite element analysis were carried out by Cook et al. (1982). A model was constructed to represent a baboon mandible containing a blade-type implant. The model consisted of a total of 476 nodal points and 600 tetrahedral and hexahedral elements. Three implant infrastructure design were studied. In addition to the implant, a molar with a periodontal membrane surrounding and a three unit fixed bridge connecting the tooth and the implant was included in the finite element model (Fig.62).

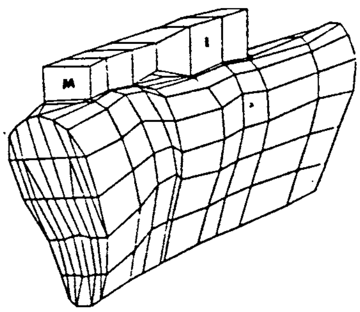


Figure 62 Representation of the three-dimensional finite element model; (M) Molar, (I) implant (Cook et al., 1982).

All implant and mandibular tissue material properties were assumed to be linearly elastic and isotropic with material elastic properties obtained from the literature. Cancellous bone representation was similar to that reported by Knoell (1977). A vertical load of 450N was applied on the model.

The effect of blade length on stress distribution around bridged aluminium oxide implants was found to be more pronounced than that around LTI carbon implants. They also observed a 50 percent increase in stresses in the cortical plates when the blade length was reduced to half size.

To investigate the effect of a resilient layer in a denture base to the mandibular bone, Aydinlik and Akay (1980) tried to construct a two-dimensional finite element model of a mandible with missing posterior teeth. The mesh consisted of 238 elements of triangles and quadrilaterals with a total of 265 nodal points. The triangles were the constant stress elements, whereas the quadrilaterals were the bilinear displacement elements with incompatible bending modes. Plane stress elastic condition was assumed, and the mandibular base and the ramus were constrained. A vertical masticatory force of 25 kg was applied to the dentures (Fig.63).

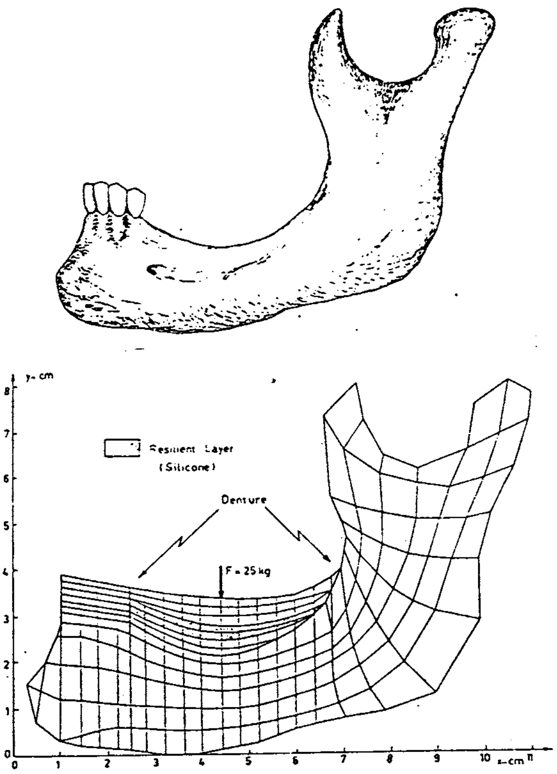


Figure 63 Finite element mesh of the mandible with a removable partial denture. (Aydinlik and Akay, 1980).

They have reported that the resilient layer acts as a "shock absorber" and redistributes the effect of a concentrated load uniformly along the base of the denture. They also have reported that the distal abutment tooth region and the retromolar region were subjected to high tensile and compressive stresses. Since the periphery of the mandibular base and ramus was not fixed against movements, these high stresses were attributed to the differences between the 2-D model and the actual gnathostomatic system.

In their study of mandibular mechanics employing current methods for assessing resistance of materials, Ferre et al. (1981, 1982) demonstrated the use of finite element method to mandibular fracture mechanics. Since the work was on review of the current methods utilized in mandibular structural mechanics, details of this simplified model was not given (Fig.64).

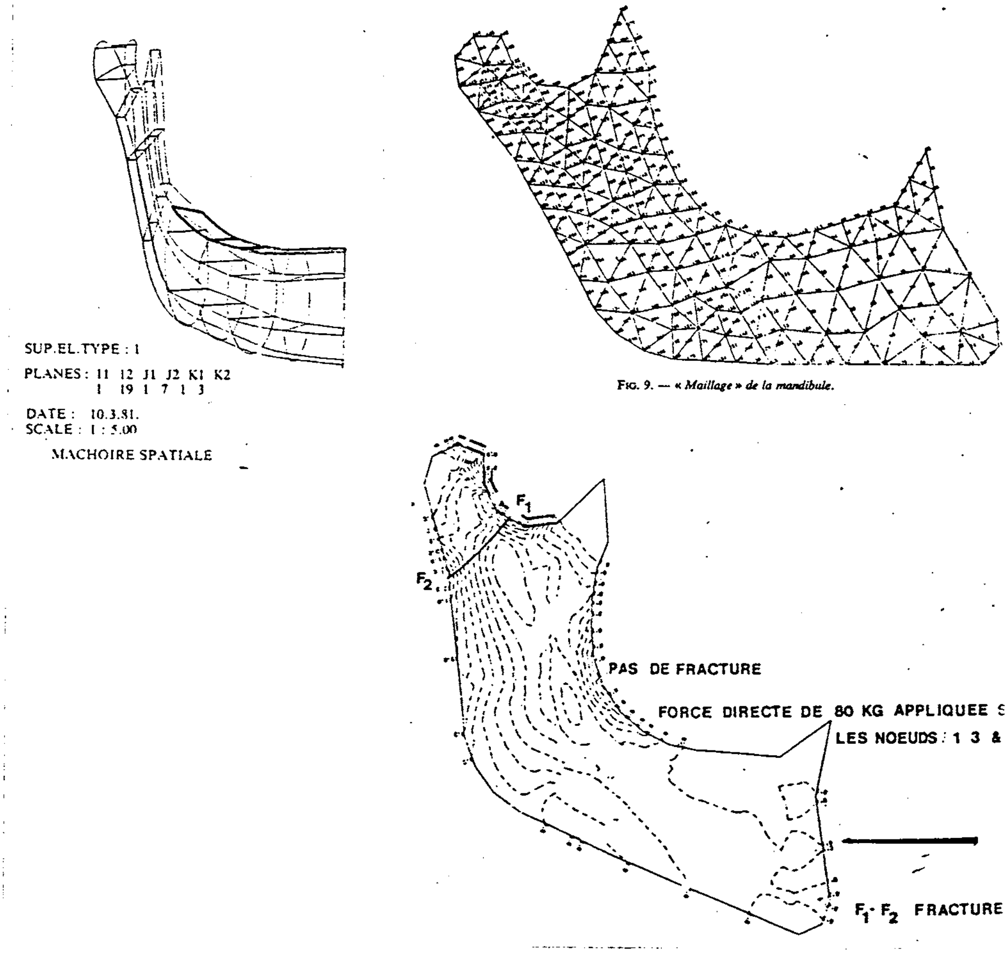


Figure 64 Finite element model of the mandible, and the fracture path given under 80 kg applied force. (Ferre et al., 1981, 1982).

The first three dimensional, solid mandible model, symmetric about the median line, was developed by Gupta et al. (1972). The model geometry was derived from physical measurements taken of a mandible. Nodal stresses and deformation of the model under the applied load were generated using the computer program "Elas" developed by Utku and Akyuz (1968).

Three different materials were assumed to be associated with the mandible model. Two of these, dentin and alveolar bone, represent the actual dental tissues; the third represents a material with finite element meshes covering both dentin and alveolar bone areas. These materials were assumed to be isotropic and linearly elastic with modulus properties as shown in Table 11 in Appendix IV. The elastic moduli of alveolar bone were assumed to be similar to those of skull bone (McElhaney et al., 1970), with 50 percent porosity. The periodontal ligament was not included in this model.

The idealized structure consisted 271 nodes and 240 solid elements (Fig.65). A total load of 445 N was uniformly distributed on the occlusal surface of the first bicuspid in the mathematical model, which was then analyzed to measure the nodal stresses and the deformation.

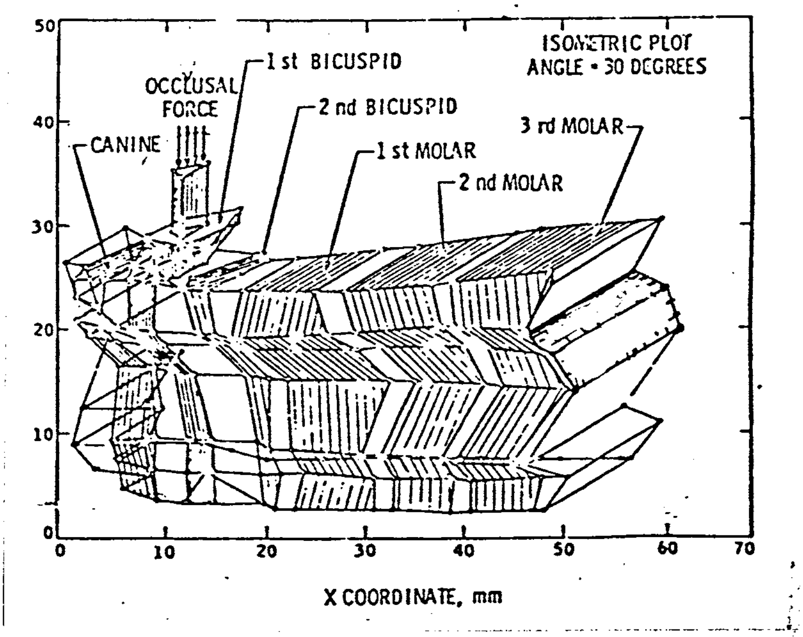


Figure 65 Three dimensional view of the first FEA mathematical model of the mandible (Gupta et al. 1972).

Constraint points of the model, unfortunately was not reported, but it is assumed that the mandibular base was chosen to be the constrained area.

The results of their numerical analysis were compared with experimental results obtained on a dried human mandible using holographic interferometry. They have reported a good agreement with holographic test results, which according to the investigators, it has established the validity of the mathematical model.

The mandible model of this preliminary work was modified by Norton et al. (1974) by replacing the first bicuspid with a blade-vent implant. The model geometry, aside from the implant, was the same as that of the prior model, which had been obtained from measurements taken from a horizontally sectioned mandible (Fig.66).

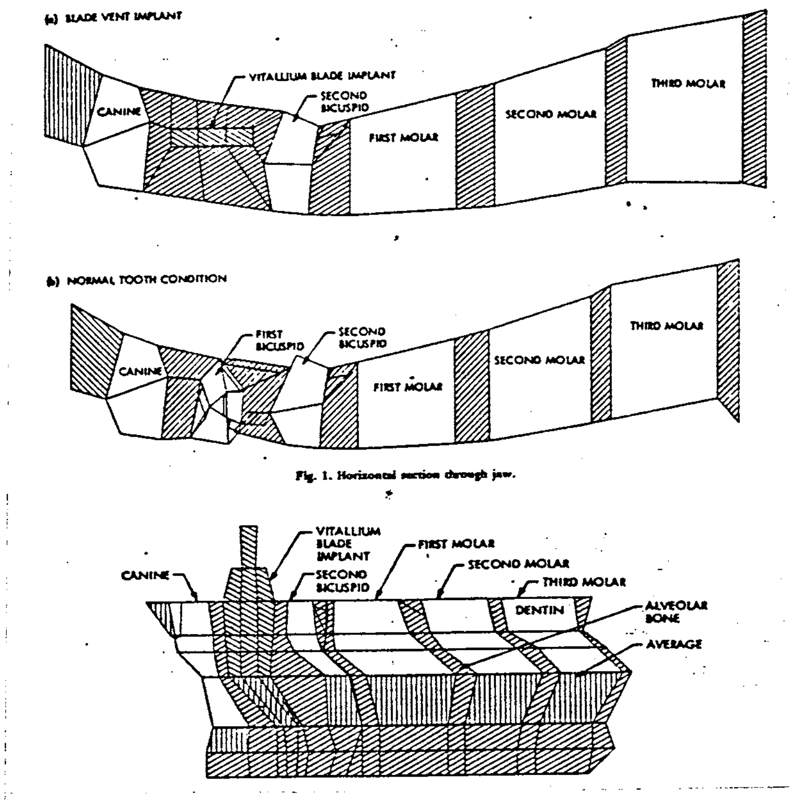


Figure 66 A model of a horizontal section of the mandible, comparing the blade vent implant with the normal tooth (Norton et al. 1974).

The loading used was a 445N vertical force on the implant. The stresses and deformations were again computed using the general purpose structural analysis program,"Elas". The finite element mesh consisted of 292 nodes and 166 elements. The boundary conditions used were complete fixity of the bottom surface. The material properties with the exception of the vitallium implant were the same as the previous work.

The results obtained were compared with the previous analysis of the normal mandible under similar loading to demonstrate procedure for evolving first order biomechanics design criteria for dental implants.

The second generation three-dimensional model of the mandible was attempted by Knoell (1977). This improved model consists of a three dimensional, finite element representation of one-half of a dried in vitro mandible symmetric about the symphysis (Fig.67). In comparison to the previous work, this model contained full mandibular dentition, including third molars and the ramus region. Even though the elements were coarse and resulting model was not refined, it is interesting to note that it was the real, first attempt to model the complete mandible.

The materials simulated in the model include dentin and cortical and cancellous bone. The properties of the cancellous bone were developed by scaling the properties of cortical bone on the basis of porosity as determined from void area measurements taken of the mandible sections. This approach was consistent with the mathematical model of trabecular bone developed by Pugh et al. (1973) which was used to characterize its mechanical behaviour. Specifically, for a cancellous bone element of the mandible model, a ratio of the section surface void area to the total element area was determined, and this ratio was used as a scaling factor on the cortical bone properties to determine those of cancellous bone. By means of this approximate scaling technique, he has defined five types of cancellous bone properties, ranging from 1 to 65 percent of cortical bone properties (Table 13, Appendix IV). In his model the specific value used for the element under consideration was that which most nearly matched the void ratio as determined above. In general, the largest values for cancellous bone properties were used in the region of the alveolus, whereas the smallest values were used in the area of the mandibular canal.

The materials were idealized as homogeneous, isotropic, linearly elastic solids. The basic material properties for dentin and cortical bone were taken from the literature (McElhaney et al., 1970; Grenoble et al., 1972; Yamada, 1970). The periodontal ligament was not included in the model development. The finite element model of the mandibular structure consists of 674 nodes and 941 elements.

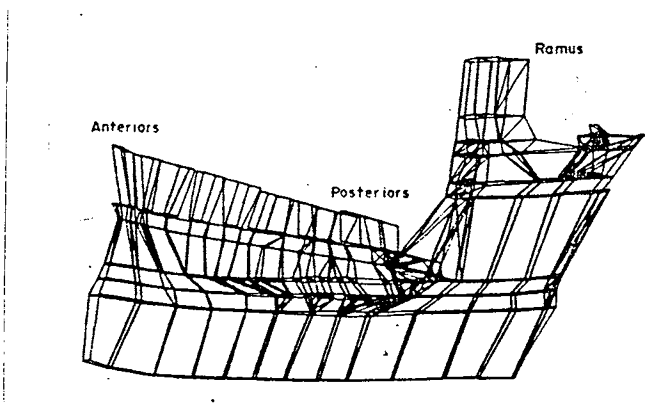


Figure 67 Finite element model of mandible (Knoell, 1977).

The elements were hexagonal, tetrahedronal, and wedge type constant strain element discretizations. The model was analyzed for several cases of vertical and horizontal static loads individually applied at the occlusal surface of both the first bicuspid and molar, respectively. The loadings were reported to consist, a vertical force of 445 N and a horizontal force of 44.5 N. The model was assumed to be fixed against displacement along the base of the mandibular canal. No other force or load was reported to be used in this model.

This 3-D model was analyzed using the Nastran program (McCormic, 1973) on the Univac 1108 computer.

In his results, the stress analysis data indicated that rapid changes in biomechanics response tend to occur in highly localized regions of major supporting hard tissues of the mandible. He further concluded that this observation highlighted the need to employ mathematical modelling techniques capable of accurately simulating the anatomy of the mandible.

The mathematical models used in these studies have been almost exclusively first-order primitive, partially completed, over constrained structural models exhibiting limited anatomical description and properties. Since structures in oral cavity are complex by nature the only useful mathematical models of the mandibular environment are the ones which represent these complexities. This improvement must include, forces acting on the mandible, realistic geometry and boundary conditions analagous to musculature support and ultimately better bone and tissue property characterization both structurally and biologically.

CHAPTER IX

MATHEMATICAL ANALYSIS

As mentioned in the introductory chapter of this thesis, most of the previous work on the mandibular distortion were carried out by dentists, whose only interest was the measurement of the degree of adduction during the various movements of the mandible. To date, no mathematical model of the mandible has been attempted to solve this or other related problem.

There were various reports on two dimensional and three dimensional modelling of the mandible. These models appear to be very remote from the actual anatomical and biomechanical conditions. However, it must be remembered that it is extremely difficult to perform complete three-dimensional mechanical analysis of the mandible with its complex configuration and structure. The use of computer is essential if the analysis would be applied to physiological problems. The theoretical analysis presented here was devised with this purpose in mind. Use has been made of the capability of the pre processors to perform repetitive calculations quickly.

The general purpose of this study was to improve the method of three dimensional modelling and to reduce the gap of biomechanics and medical science by utilizing the model in an important maxillofacial problem; namely "functional distortion of the mandible".

It is this author's belief that very complicated biomechanical problems should start with the most simplified form of geometry to reduce the effect of the certain parameters to minimize the problem.

Although most investigators do agree that, in view of all the simplifying assumptions, only the trend of the results produced by their model could be important, absolute numbers have often in the past presented, discussed and passed on. As a consequence, misinterpretation of the data has frequently occurred.

The results of simplified, general models should be stated in qualitative terms, and in this case the question of validity too, is of a qualitative nature. The validity of a model can be addressed in several ways. Very little attention has been devoted to experimental verification in finite element modelling areas, and where it was, the experiments were not always satisfactory.

In present work, after the investigations of the mandibular distortion by utilizing the developed mandible model, published experimental data was compared to prove the validity of the model.

This work was initiated to model an edentulous human mandible by using high order linear, three dimensional solid elements. Finally this model was utilized to analyze the functional distortion of the mandible during normal opening, protrusion and under the clenching conditions.

The analysis was carried out in the following stages:

i) Modelling of the mandible using digitizer and computer aided design software.

ii) The development of CAD/Finite Element Analysis modelling sequence.

iii) Rationalisation of input data: Mechanical properties and the muscle forces acting on the mandible during opening, protrusion and clenching.

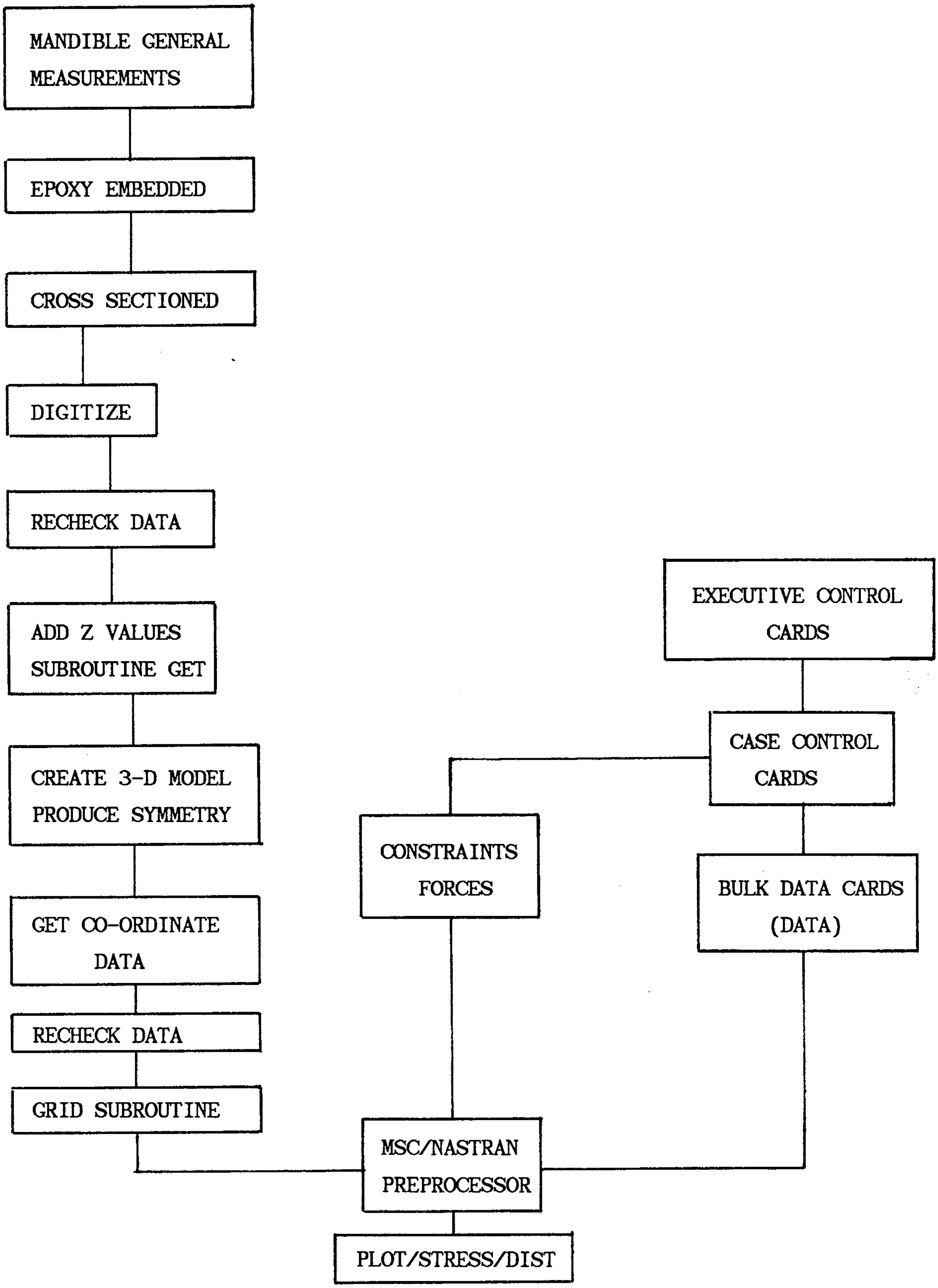
iv) The muscle forces required to balance the external system were found and from these, the joint forces were calculated.

v) FEA using MSC/Nastran programmes.

9.1 Method of Analysis

In this work, important data on anatomical, structural, functional and materials aspects of the mandible were obtained from various sources. These data were later applied to the three dimensional finite element modelling of the mandible.

The model geometry was derived from physical measurements taken of a mandible. One half of a representative human mandible was embedded in epoxy and cut vertically with a diamond saw into 30 sections. The other half was cut horizontally into 17 sections. Horizontal sections were utilized on METSAP preprocessor preliminary analysis. While vertical cross sections were digitized according to finite element modelling order, then sent to MSC/NASTRAN preprocessor to carry out analysis of the stresses and displacements under the applied functional forces. The summary of this procedure is given in the following flowchart (pp.151).


MANDIBLE GENERAL MEASUREMENTS
EPOXY EMBEDDED
CROSS SECTIONED
DIGITIZE
RECHECK DATA
ADD Z VALUES SUBROUTINE GET
CREATE 3-D MODEL PRODUCE SYMMETRY
GET CO-ORDINATE DATA
RECHECK DATA
GRID SUBROUTINE
EXECUTIVE CONTROL CARDS
CASE CONTROL CARDS
CONSTRAINTS FORCES
BULK DATA CARDS (DATA)
MSC/NASTRAN PREPROCESSOR
PLOT/STRESS/DIST

9.2 Methods and Equipment Used

9.2.1. MSC/Nastran

MSC/Nastran is a proprietary version of Nastran, the NASA sponsored computer program for structural analysis by the finite element method. Today there are three versions of Nastran. The official version is distributed by NASA through Cosmic. One version is distributed by Sperry Univac. The version used in this work is the MSC/Nastran. This is the version maintained and continuously developed by McNeal-Schwendler Corporation.

MSC/Nastran is a self-contained system and cannot be described as a suite of small programs. It is a large-scale general-purpose computer program that can solve linear static and dynamic as well as material and geometric nonlinear problems. Nastran has a number of so-called rigid formats which are prepackaged programs for static, dynamic and buckling analyses. For problems that cannot be solved by these rigid formats the user may write his own program performing the appropriate computational steps.

9.2.2. Metsap

Metsap is a UNSW modified SAP IV program which is a general three dimensional, linear, static and dynamic finite element program. A pre-and-post processor graphics subroutine was added to the SAP IV version. Plotting of the undeformed and deformed structure and the vibration mode shapes of a structure are also possible.

9.2.3. Digitizer

Digitization is a process where the graphic material in the form of designs, tracings, layouts, patterns etc. could be converted into sets of x-y co-ordinates.
Summagraphics ID, with a builtin microprocessor providing the computing capability was used for digitizing the cross sections of the mandible.

9.2.4 Palette CAD-Database Software:

Palette is a software package supplied by Palette Systems Ltd. Australia, consisting of a number of separate but highly integrated software modules for use in the fields of computer-aided design and spatial databases. In the CAD field, Palette provides complete ready-to-use software modules for both 2-D design and 3-D modelling, while in the spatial database field, it provides sophisticated tools for the development of integrated systems. It can run on all VAX/VMS configurations and supports a wide range of terminals, plotters, digitizers and scanners. The 3-D modelling module allows the construction of a model in three-dimensional space, with simultaneous viewing and interaction occurring in multiple projections. The 3-D module is integrated with the 2-D Drafting module, so that two dimensional sections defined in the 2-D module can be used in the construction of 3-D models, while projections generated in the 3-D module can be deposited into the 2-D module for detailing and dimensioning.

Two basic mechanisms are provided for integrating Palette with other programs, namely, file interfacing and the Independent Program Interface. File interfacing permits the two-way transfer of information between Palette and other programs by means of formatted ASCII files, while the Independent Program Interface uses sophisticated multi-tasking to provide truly interactive interfacing. Palette provides a library of subroutines which could be linked into their application programs and handle all the necessary intertask synchronization and cummunication. The main benefit of the Independent Program Interface is its open-ended nature. By allowing control of Palette to go completely outside into an independent program, without imposing any limitations on the independent program.

9.4 Method of Modelling

Three different 3-D mandible models have been developed to analyse the mandibular distortion. The preliminary model was a primitive model prepeared using simplified "U" shape geometry. With an exception of the mandibular general dimensions, none of the available anatomic data was utilised. The model consisted 52 nodal points and 12 elements (Fig.68). This preliminary model was analysed using Metsap FEA. After this preliminary distortion analysis, the real anatomical model was attempted using the data obtained from a cross sectioned mandible. This first realistic model was homogeneous isotropic and similar to the ones previously reported. The second one was a more advanced model consisting both cortical and cancellous bone regions.

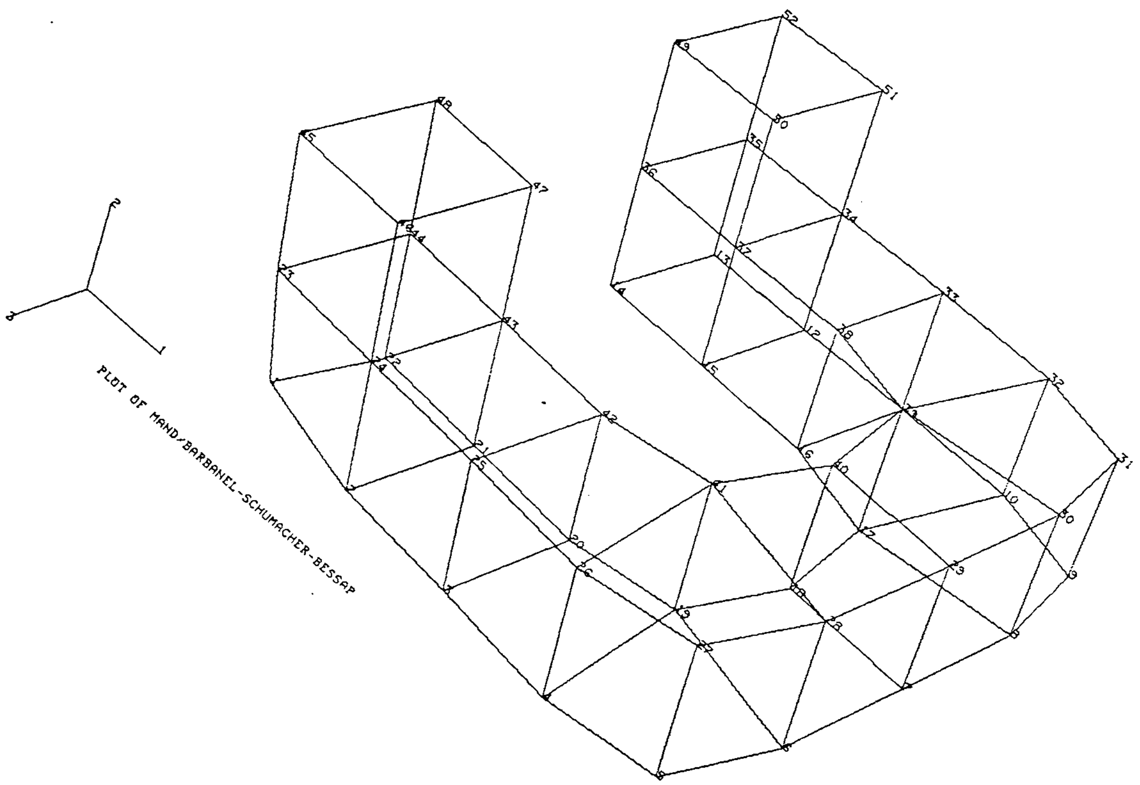


Figure 68 The preliminary "U" model.

9.4.1 First Stage Modelling

In this model, the structure of the mandible was considered to be symmetric, and the development of the model was attempted on one half of the mandible. Full mandible was embedded in epoxy and was cut vertically along the symphysis. One half of this model was cross sectioned horizontally and was utilized in the analytical representation of the first stage modelling (Fig.69, 70).

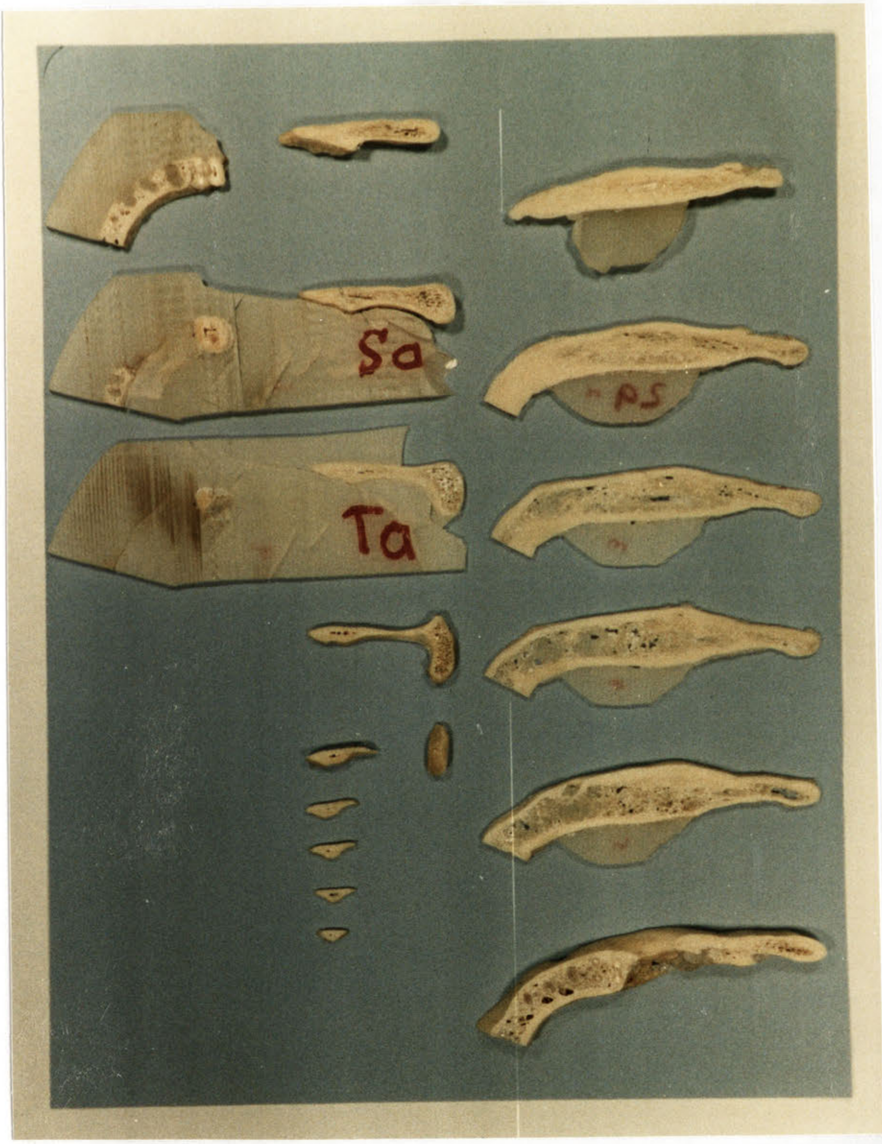


Figure 69 Horizontal cross sections used in the first stage model development.

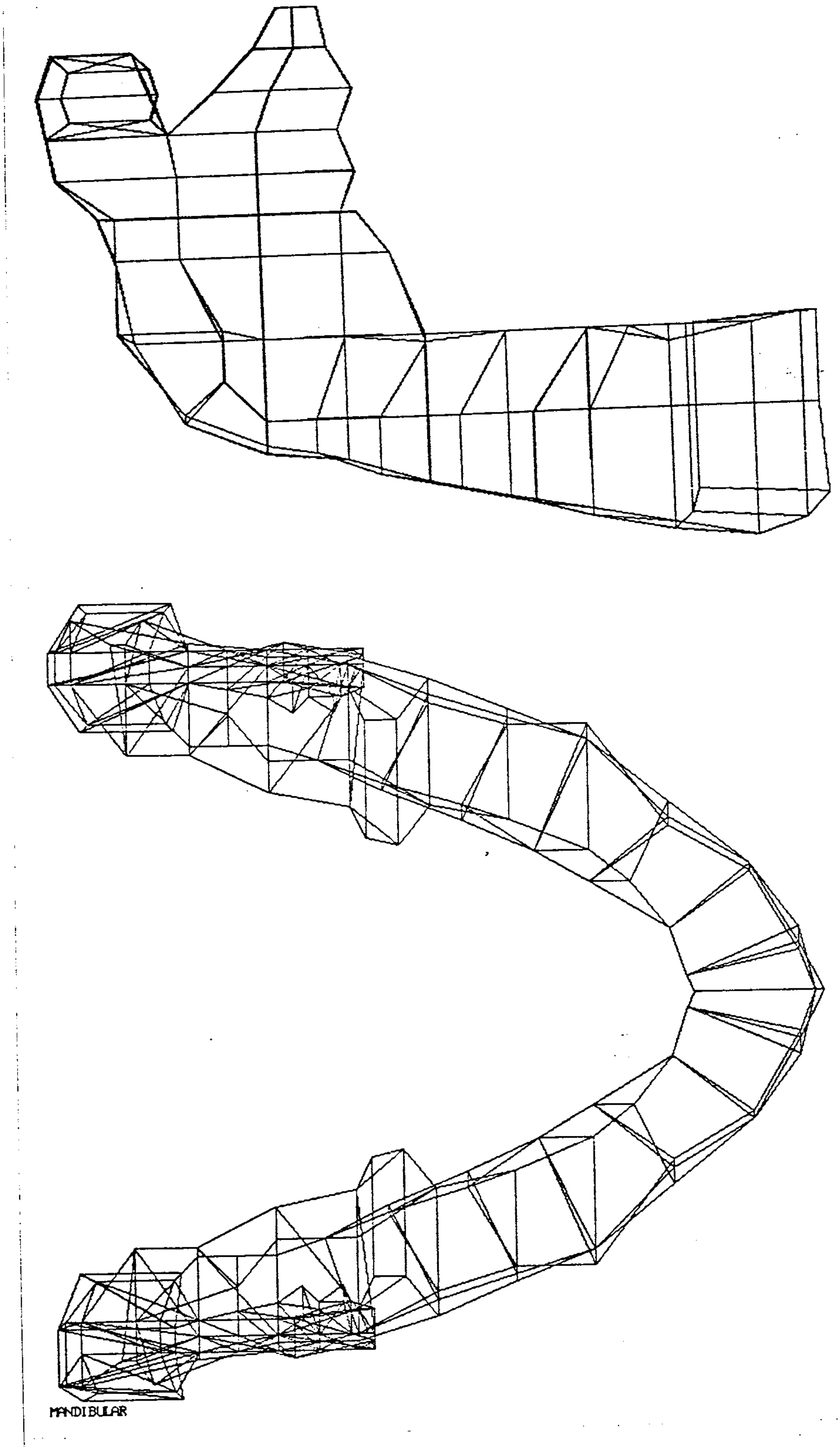


Figure 70 General nodal modelling sections for FEA .

The finite element model was then constructed using measured nodal values from these cross sections. The finite element model of the first complete mandibular structure shown in Fig.71 consists of 272 nodes and 126 solid hexagonal (8 nodes), and pentahedral (wedge type) elements.

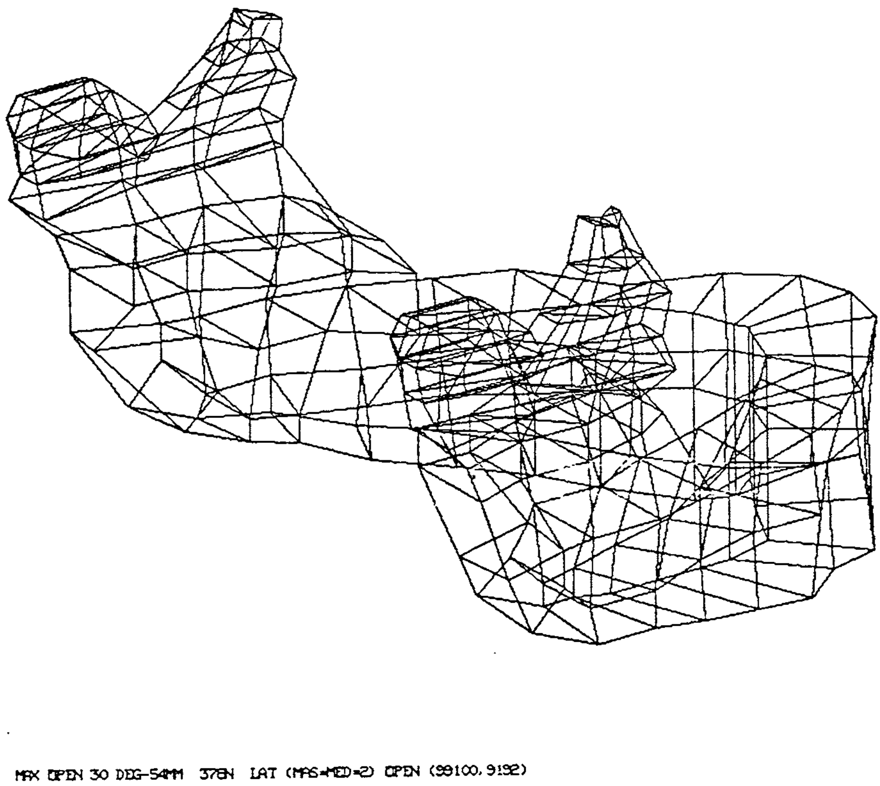


Figure 71 First stage finite element model of the mandible.

To fix the mandible in space we need to fix out the six rigid body motions by defining points of zero displacement. If no load is applied at that point, the reaction to the fixity is zero and it will not effect the stresses in the mandible. If the total force at the nodes not constrained is not zero, then the reaction will be the force to cause equilibrium. Hence we can apply a force at the constrained nodes by making the forces at other nodes out of equilibrium to the required value. For this analysis three nodes on the symmetry plane were restrained (Fig.72). The model was completely free to deform removing the approximation caused by over restraining in previous models (such as Norton et al.,1974; Knoell, 1977).

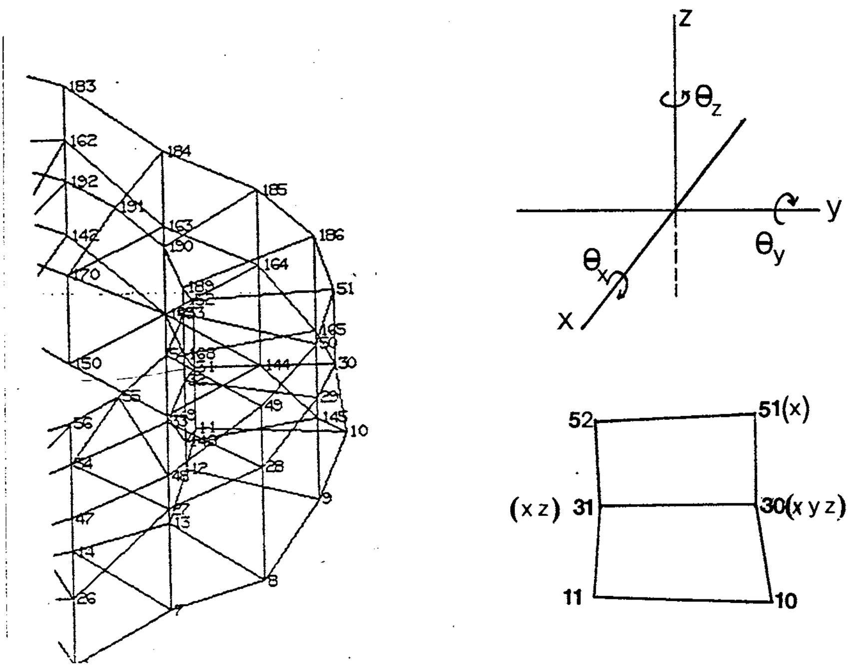


Figure 72 Finite element model of the constrained section (a), and Possible rigid body motions (b).

In the figure 72, fixing node 30 in all directions prevents translational rigid body motion in all the coordinate directions. Fixing the x displacement at node 31 prevents rotation about the z axis; fixing the z displacement at node 31 prevents rotation about the x axis; and fixing x displacement at node 51 prevents rotation about the y axis.

In this first stage model mandible was assumed to be homogeneous, isotropic and Young's moduli was used as 1.03×10^4 N.mm^{-2} while Poissons ratio was 0.3 .

Clenching, maximum opening and protrusion were analyzed in the distortion analysis.

9.4.2 Second Stage Modelling

Advanced, second stage modelling was attempted with an mandible embedded into the epoxy with a wire, which was later utilized as a centreline marker (Fig.73).

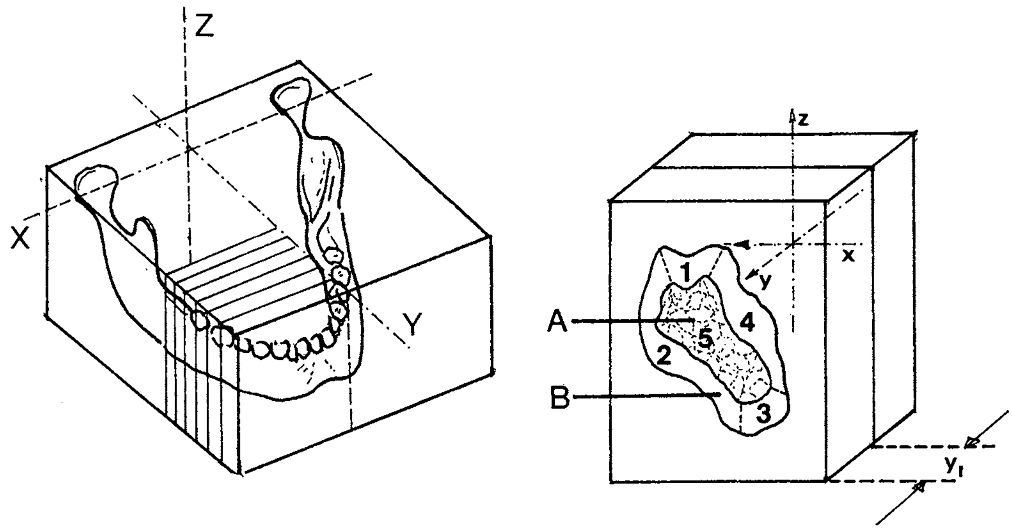


Figure 73 Cross sectioning of the mandible and the element divisions.
A) Cancellous bone, B) Cortical bone.

Mandible was cross sectioned vertically using isomet diamond blade cutter to minimize the loss of the bone. Cross sectioned areas (Fig.74) then were digitized by using Summagraphics ID digitizer and specifically developed data producing sequence.

Each cross section of the bone was divided into five sections, outer four representing the cortical bone and the inner one the cancellous bone (Fig.73).

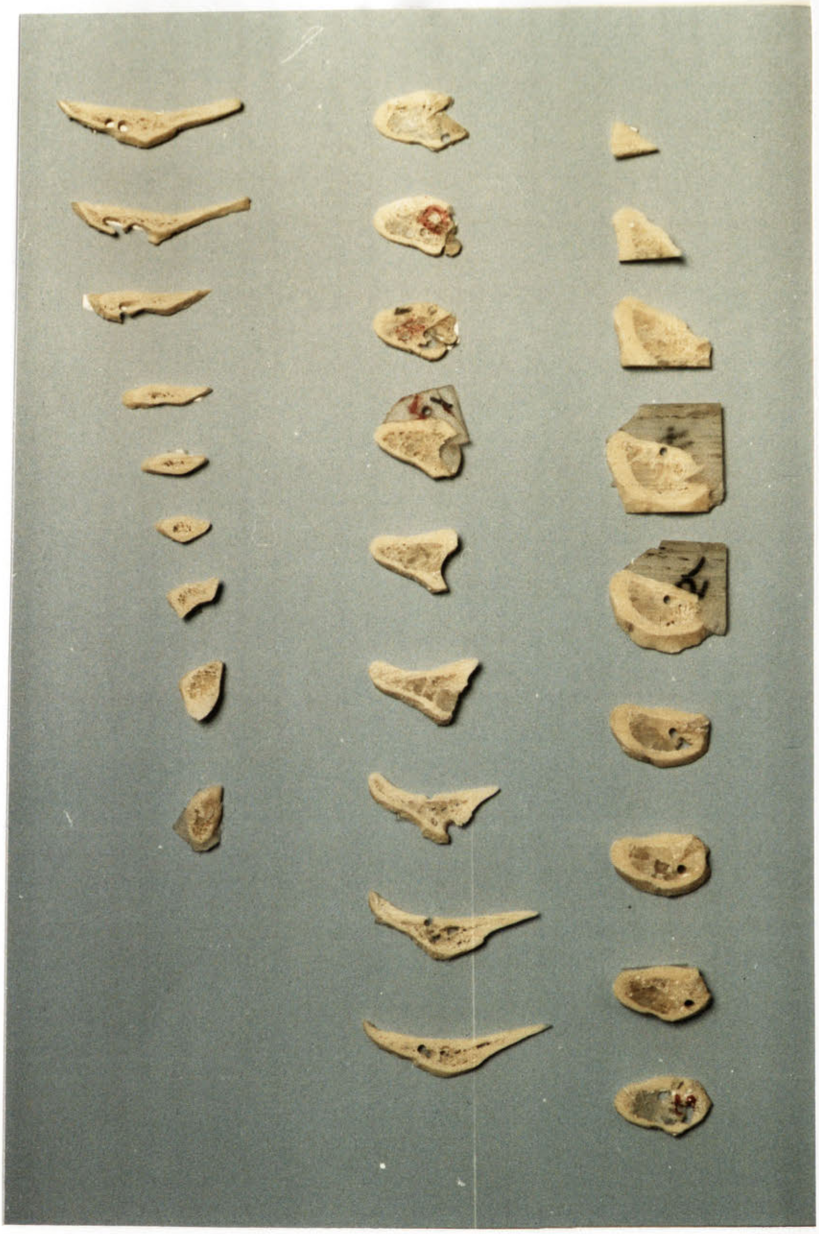


Figure 74 The photograph of the cross sections of the one half of the mandible.

To model this composite, a number sequencing method was developed to match the finite element input data sequence (Fig.75).

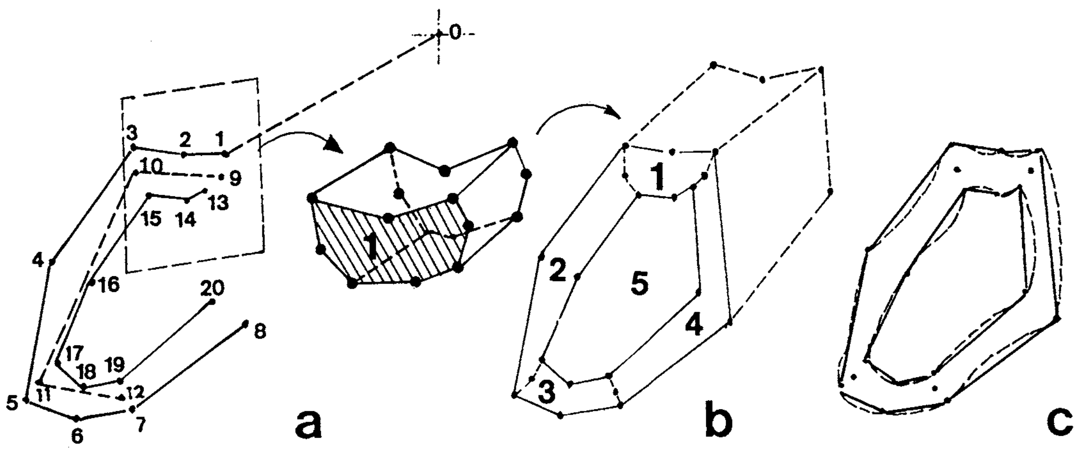


Figure 75 Nodal point generation using digitizer.

a) digitizing sequence and wire frame diagram,

b) element perimeter and midside nodes on cross section,

c) element perimeter superimposed on real cross section.

During this 2-D modelling, the direction of the number sequencing is very important. Wrong sequence can change the input data for the preceding finite element work. Once the 2-D model is completed, the cross sections can be joined using the known thickness (y) values, obtained during the cross sectioning of the mandible. A small programme was utilized during this joining process.

The three dimensional model (Fig.76)then was completed using the Palette system and once the obvious geometric faults identified and corrected, general co-ordinate data was generated. This generated data could then be compared with the bulk mandibular measurement's and if the co-ordinate data is satisfactory, it can be sent to MSC/Nastran preprocessor for analysis. To run the Nastran finite element program, force and material properties data should also be included in the bulk data cards.

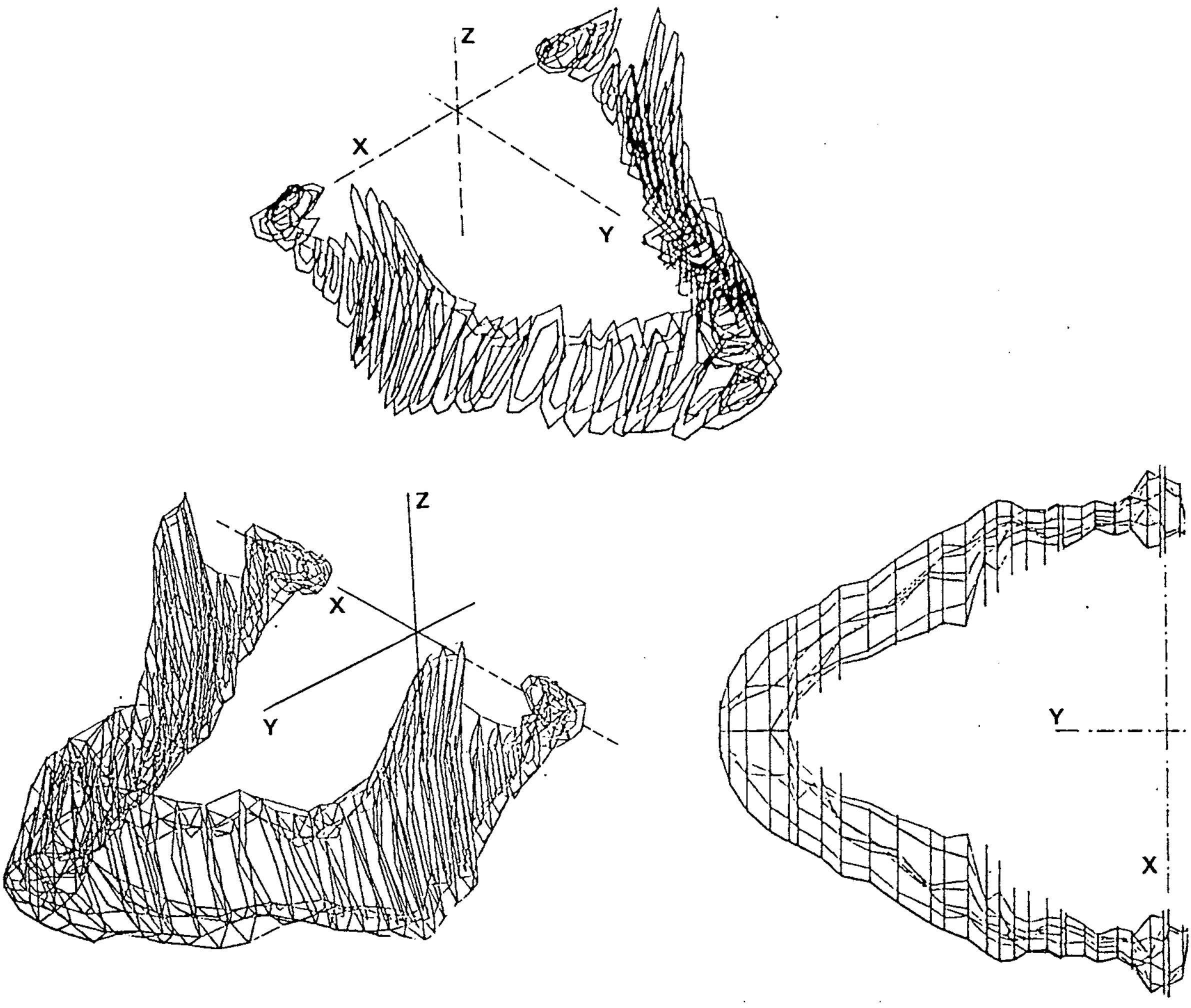


Figure 76 The second stage model showing different views of the mandible.

In this model the materials were idealized as homogeneous, isotropic and linearly elastic. The basic material properties for cortical bone and cancellous bone were taken from the previously published values.

The model was fixed at the symphysis similar to the previously developed model (Fig.77).

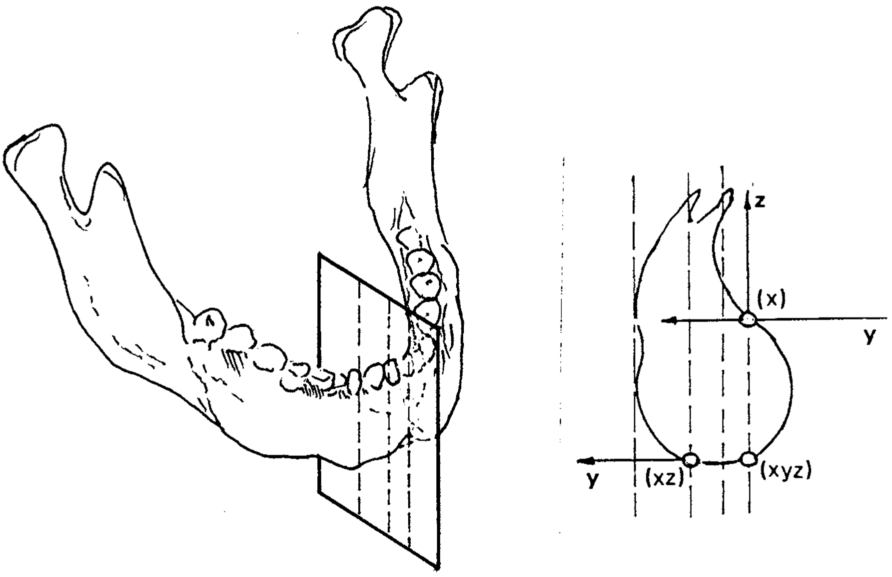


Figure 77 Cross sectional area at the symphysis showing the constraint points.

To simplify the procedure an edentulous human mandible was modelled. This second stage model consisted 258 hexagonal (20 node) and pentahedral solid elements and 1106 nodes.

9.5 Mechanical Properties of the Mandibular Bone

There seems to be no published data specifically dealing with the mechanical properties of the mandibular bone. But the behaviour of bone as structural element in human body has motivated a large number of investigators into the mechanical properties of bone, extensive and comprehensive reviews by McElhaney et al. (1970), Evans (1973), Katz and Mow (1973), Reilly and Burstein (1975), and Burstein et al. (1976) are noteworthy. More recently reviews on the overall implications of selected published data with respect to current understanding of elastic and viscoelastic deformation were given by Bonfield (1984), Lakes and Katz (1984) for cortical bone, and by Van Audekercke and Martens (1984) for cancellous bone.

Generally speaking bone is a heterogeneous, anisotropic, multiphase composite. There are several proposed classifications of bone and one allows for five kinds of bone. These are: woven, bundle, fine cancellous, cortical and cancellous bone. In mammals, the first three are mainly temporary tissues, associated with the growing skeleton to be replaced by the other two types of bone, cortical and cancellous bone in the adult.

Compact or cortical bone found in mature birds and mammals is made up almost entirely of dense, fine fibred bone. In this bone the vascular channels are very narrow so that to the naked eye the tissue appears to be made up of hard bone and nothing else.

Cancellous (trabecular, spongy) bone is structurally open cell foam which is present at the epiphyseal and metaphyseal region of long bones and within the cortical confinements of flat and short bones. This bone in the adult mammalian skeleton is similar, in its fine structure, to compact bone but is sponge like in appearance, and is made up of trabeculae of bone tissue. Cancellous bone is continuous with the inner surface of the cortical shell and presents a three-dimensional lattice composed of plates and columns of bone. The mechanical properties of cancellous bone have been studied less thoroughly than those of cortical bone.

In present study the mandible is assumed to consist cortical and cancellous bones. Alveolar ridge, teeth, dentin, enamel and periodontal membrane properties were not utilized in the edentulous model.

The various published values for the mechanical properties of these materials are given in Appendix IV.

In present study various mechanical properties values of cortical and cancellous bones from Black and Korostoff (1973), Gupta et al. (1973), Widera et al. (1976), Knoell (1977), Svensson et al. (1977, 1980), Weinstein et al. (1980), Lavernia et al. (1981), and Cook et al. (1982) were utilized at different boundary conditions.

9.6 Forces Acting on the Mandible

As mentioned in previous chapters, in the early biomechanical models of the mandible, the forces exerted by some important muscles were not considered. The direction of the bite force and the direction and magnitude of muscle forces were largely unknown. In most cases the mandible was assumed to be made of homogeneous and isotropic rigid material with uniform cross section and with all loads and reactions applied in the same plane.

In general terms, any mathematical model in biomechanics, especially the mandible requires sets of information.

i) The magnitude and directions of each muscle force applied according to mandibular movement;

ii) The magnitude and direction of the bite force (if applied);

iii) The magnitude and direction of the reaction force; and

iv) The lengths of the moment arms of the bite and/or muscle forces.

The muscle forces considered in this analysis were selected from Mainland and Hiltz (1934), Barbanel (1972), Schumacher (1961) and Pruim et al. (1980). All the forces were assumed to be symmetrical with respect to the mid-line and to have an equal magnitude on the right and left side of the mandible.The forces exerted by contracting muscles were represented by vectors. Similar assumptions were made by previous workers (Paul, 1967; Morrison, 1967; Paulson, 1973). This assumption is reasonably true when the muscle is homogeneous and acts as a whole.

The direction of these vectors can be defined by the connecting lines between the centroids of the origins and the insertions of the muscles. Descriptions of these areas of attachments were derived from the anatomical literature (Schumacher, 1961; Mainland and Hiltz, 1934; Sicher and DuBrul, 1975; Hawthorne, 1969; Hylander, 1978) and three dimensional observations and measurements on different skulls.

Physiological cross sectional area data available (Schumacher, 1961 and Pruim et al., 1980) were used to assume the forces of the temporalis, lateral pterygoid, digastrics and medial pterygoid muscles.

The bite force was assumed to be around 700 to 800 N according to Pruim et al. (1978, 1980)'s measured and calculated values (Appendix II).

In this study three mandibular movements were investigated: opening of mandible, protrusion and clenching.

For the finite element model to be valid the applied forces must be in equilibrium so that the reactions at the restrained nodes in figure 77 will be zero. It was therefore necessary to define physical conditions during various mandibular movements to reduce the complex force systems to a form which allowed the definition of valid load cases for the finite element model.

9.6.1 Clenching

When analysing bite forces or clench (when the jaws are pressed firmly together without bolus), the mandible may be considered to be in static equilibrium so that the sum of the forces on it and the sum of the torques generated by the forces about an arbitrary but fixed point in space are zero. This leads to six linear equations. The coordinate system to be used is shown in Fig.78 .

If F_n is the force of the n th muscle applied at a_n, B_k the k th bite force applied at b_k and J_q the q th joint force applied at c_q , the conditions for equilibrium are:

$$\Sigma F_n + \Sigma B_k + \Sigma J_q = 0 \quad \text{and} \qquad 9.1$$

$$\Sigma a_n F_n + \Sigma b_k B_k + \Sigma c_q . J_q = 0 \qquad 9.2$$

N, K and Q are the number of muscles, the number of bite forces, and the number of joint forces respectively. Moments are computed about the mid point of the line segment joining the centres of condyles when the mandible is in intercuspal position.

The sum of the z components of the applied forces:

$$\Sigma F_n \cos \gamma_n - \Sigma B_k \cos \gamma_k - \Sigma J_q \cos \gamma_q = 0 \qquad 9.3$$

The sum of the y components:

$$-\Sigma F_n \cos \alpha_n + \Sigma B_k \cos \alpha_k + \Sigma J_q \cos \alpha_k = 0 \qquad 9.4$$

The sum of the x components:

$$\Sigma F_n \cos \beta_n + \Sigma B_k \cos \beta_k + \Sigma J_q \cos \beta_k = 0 \qquad 9.5$$

There are, however, at least eleven variables, composed of nine jaw muscles plus the two joint reactions (Fig.78). The equations of equilibrium for a given bite force will not yield a unique solution. There are infinitely many combinations of tensions and joint reactions that will produce a given bite force on a specified tooth. If however , we require that some function of the tensions be minimized , it may be possible to reject all but one of the solutions. By considering particular loading condition, further "conditional" equations can be introduced allowing a full solution of the system.

According to Wood (1986), the medial pterygoid muscle activity during an intercuspal clench appears to contribute to the force generated between the teeth. The lateral components of masseter on temporal muscles that are also active during clenching must also be taken into account. Any resulting distortions of the mandible and interocclusal forces thus represent the combined action of several major muscle groups acting in concert.

For this present work, the bite force (clench force) was directed at an angle of about 85 degrees to the averaged occlusal plane. Each bite force was equally divided between two teeth symmetrically placed with respect to the midsagittal plane, thus simulating Pruim et al.'s (1980) experiment.

During clenching all muscles were assumed to be active. Calculated muscle forces are given in Table V and Appendix V. The reaction forces were assumed to be acting at the centre of the condyles with an angle of 82 degrees to occlusal plane. The method of muscle force and reactive force calculations on YZ plane were based on published data (Appendix II and III), which included some 2-D muscle angles, directions and magnitudes. In the absence of published data on ZX and YX planes measurements were carried out on cadavers and skulls. The data were analysed by utilizing the 3-D mechanics. The equilibrium force system equations, the methodology and the results are given in Appendix V (Table 18).

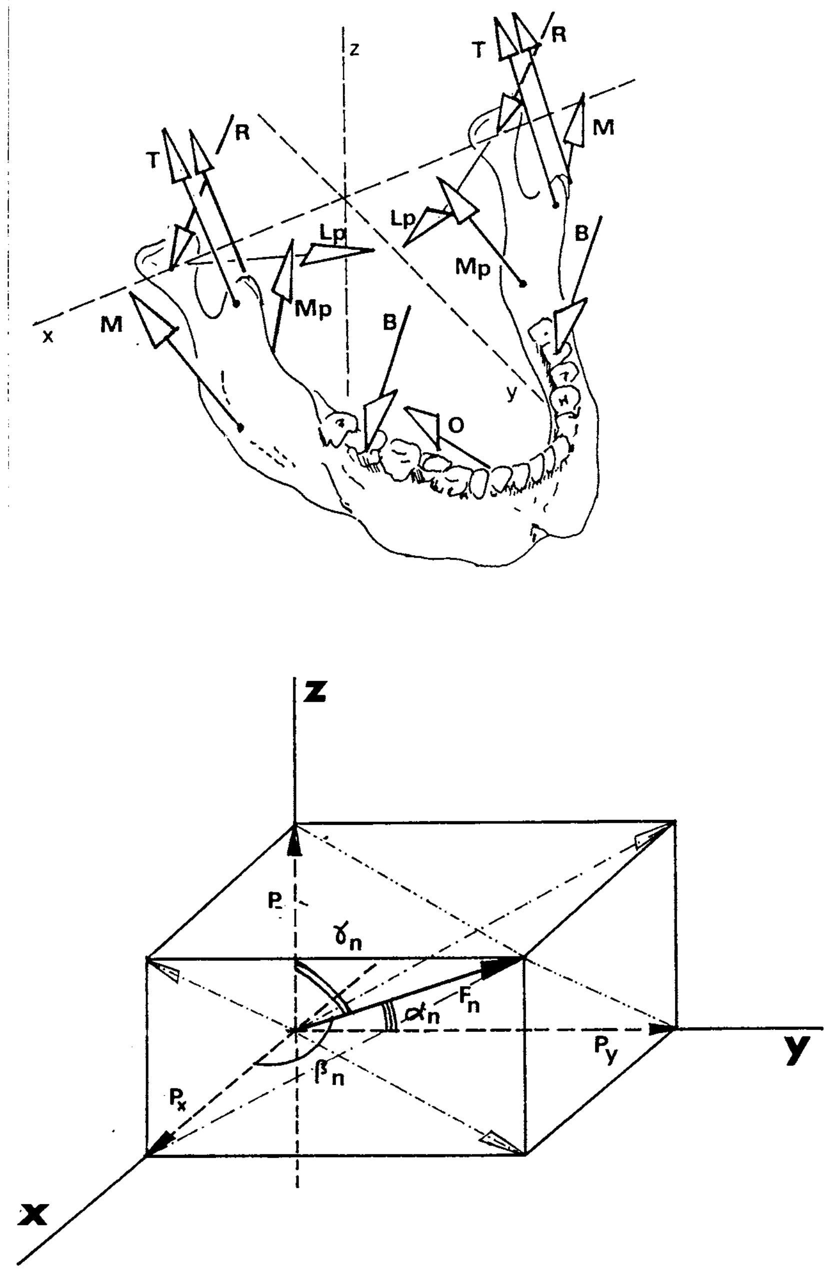


Figure 78 Applied forces and coordinate system during clenching.

TABLE V

Calculated muscle forces, and joint reaction force and bite force magnitudes (N) during CLENCH

	Px	Py	Pz	P
MASSETER A	141.9	-53.6	+304.3	340.0
MASSETER B	-141.9	-53.6	304.3	340.0
TEMPORALIS A[1]	50.6	+102.9	238.1	264.3
TEMPORALIS A[2]	50.6	+102.9	238.1	264.3
TEMPORALIS B[1]	-50.6	102.9	238.1	264.3
TEMPORALIS B[2]	-50.6	102.9	238.1	264.3
MEDIAL PTERY.A	-82.9	-29.7	170.0	191.4
MEDIAL PTERY.B	82.9	-29.7	170.0	191.4
LATERAL PTERY.A	-215.4	-307.6	-43.2	378.0
LATERAL PTERY.B	215.4	-307.6	-43.2	378.0
OPENERS	0.0	+133.2	-79.3	155.0
JOINT REACTION A	0.0	63.8	-467.6	471.9
JOINT REACTION B	0.0	63.8	-467.6	471.9
BITE FORCE A	0.0	54.7	-400.0	403.7
BITE FORCE B	0.0	54.7	-400.0	403.7

9.6.2 Opening

At its simplest, the jaw is opened by a moment couple consisting of the anterior bellies of digastric and the lower heads of the lateral pterygoids. If some of the adductor muscles "pay out" their length while maintaining some "tension",the condyle will remain in contact with the articulator eminence until,and if, it passes over the eminence.

During wide opening, masseters, and medial pterygoid muscles were assumed to "pay out" their length and guide the opening movements. Lateral pterygoid and openers muscles were utilized as main depressors (Moyers, 1950; Jarabak, 1957; Carlsoo, 1958; Munro and Basmajian, 1971; Lehr et al.,1971). Opener muscle forces were postulated according to the published physiological cross sectional area values, which included digastrics, mylohyoid and geniohyoid muscles.

The ratio of Masseter to Medial Pterygoid muscle forces was assumed to be (2:1), (Carlsoo, 1952; Schumacher,1961). The physiological cross sectional area ratios of various workers, if considered with the muscle tension factors, could be accepted to be quite reasonable. (Appendix III).

Considering the opening movement and the angle of the masseter and medial pterygoid muscles during wide opening (Fig.79). Due to the right angle created by these two muscles, the horizontal force components P_{My} and P_{MPy}, can be assumed to be zero or very close to zero. By reducing these components we can obtain the equilibrium by calculating the rest of the components of these two unknown muscles. These assumptions are based on the cephalometric measurements and some electromyographic studies. Addition to the above mentioned conditions. All opener muscles are assumed to be active at all time, i.e. no antagonistic muscle action.

Antagonistic muscle action is assumed to occur only for short periods of the opening cycle where activity in one muscle group declines as activity in another group increases. Such antagonistic action can not be gauged from the equilibrium equations and as yet there is no reliable method of measuring muscle force directly under conditions of contraction where muscle tension, length and velocity of shortening are changing.

Calculated and measured values of the forces during opening are given in Table VI.

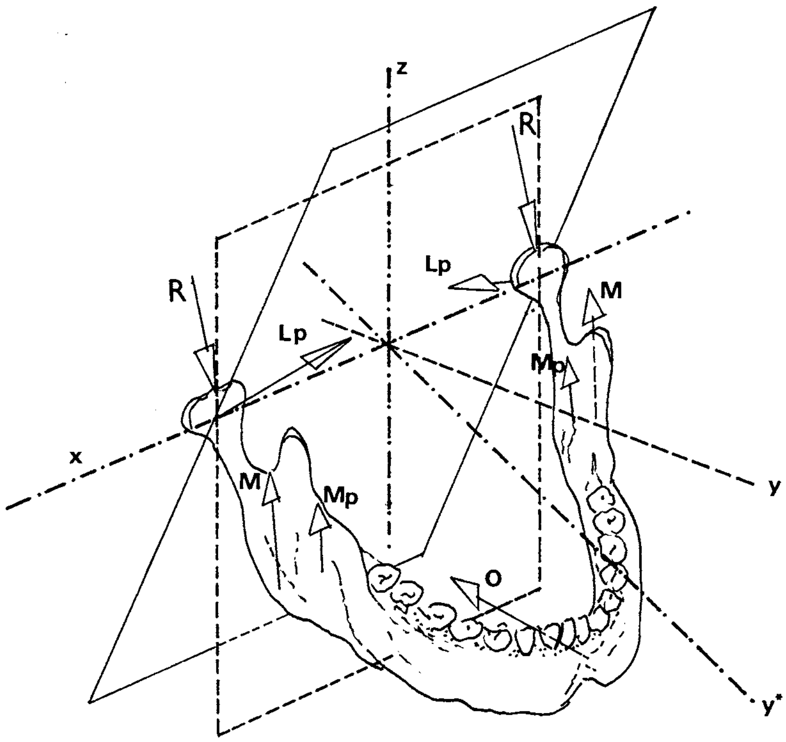


Figure 79 Applied muscle forces during wide opening.

TABLE VI

Calculated muscle forces and joint reaction force magnitudes (N) during WIDE OPENING

	Px	Py	Pz	$\overline{P}$
MASSETER A	44.7	0.0	121.2	129.1
MASSETER B	-44.7	0.0	121.2	129.1
MEDIAL PTERY.A	-24.5	0.0	60.6	65.3
MEDIAL PTERY.B	-24.5	0.0	60.6	65.3
LATERAL PTERY.A	-197.2	-281.6	156.1	377.6
LATERAL PTERY.B	197.2	-281.6	156.1	377.6
OPENERS	0.0	+135.6	-75.1	155.0
JOINT REACTION A	0.0	213.0	-300.3	368.6
JOINT REACTION B	0.0	213.0	-300.3	368.6

9.6.3 Protrusion

During the protraction of the mandible without occlusal contact, masseter was assumed to be silent while medial pterygoid was the only active elevator guiding the movement (Moyers, 1950; Carlsoo, 1952)(Fig.80). Again the horizontal component of the medial pterygoid was zero, and the other components were calculated accordingly (Table VII).

Some workers (Baragar and Osborn, 1984) assume temporamandibular ligament is active during wide open and protrusion movements. During these movements , temporomandibular ligament might be helping,on the guidence and the degree of opening of the elevator muscles. Some of the force could be distributed to this ligament during these movements. According to geometrical data the angle of the temporomandibular ligament to condyle center is similar or very close to the angle of the masseter or medial pterygoid to the condyle center during the wide open and protrusion.

These assumptions might not drastically change the final results during the flexion but might be more realistic.

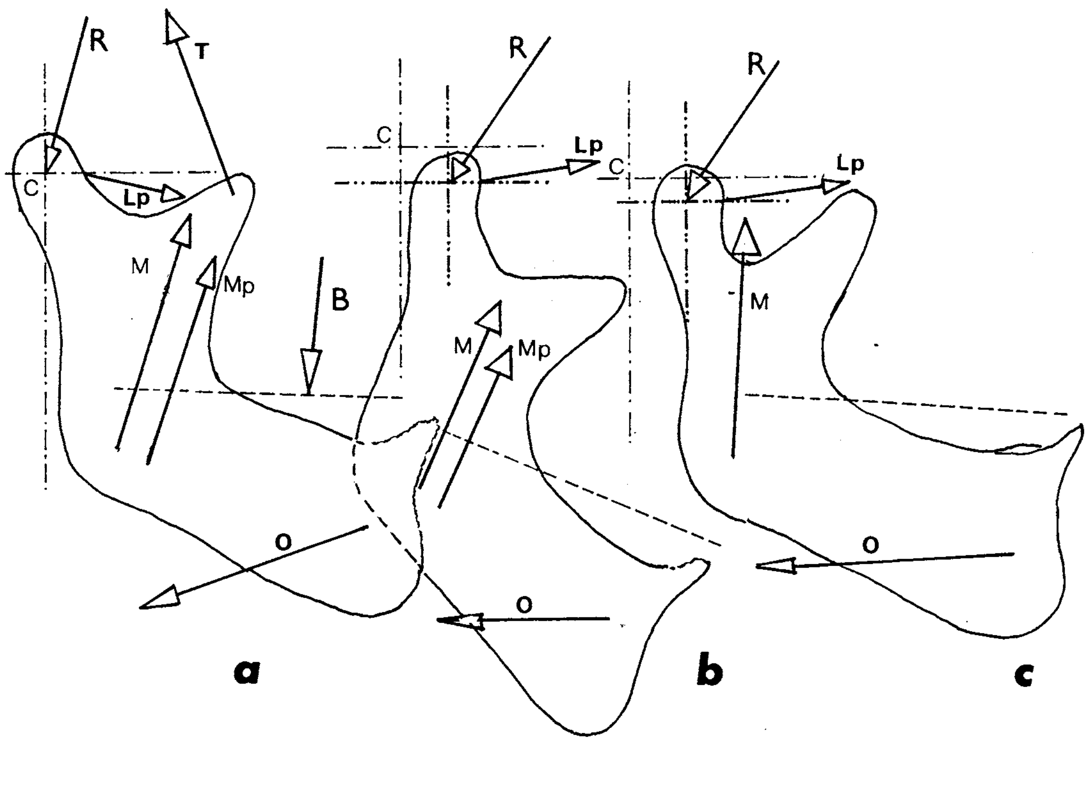


Figure 80 Lateral projection of applied forces during, a) Clench, b) Open, and c) Protrusion.

TABLE VII

Calculated muscle forces and joint reaction force magnitudes (N) during PROTRUSION.

	Px	Py	Pz	$\overline{P}$
MEDIAL PTERY. A	-74.4	0.0	181.8	196.4
MEDIAL PTERY. B	74.4	0.0	181.8	196.4
LATERAL PTERY. A	-197.2	-281.6	156.1	377.5
LATERAL PTERY. B	197.2	-281.6	156.1	377.5
OPENERS	0.0	135.6	-75.1	155.0
JOINT REACTION A	0.0	213.8	-300.3	368.6
JOINT REACTION B	0.0	213.8	-300.3	368.6

CHAPTER X

RESULTS AND DISCUSSIONS

The early primitive model prepared using Barbanel (1981) muscle values showed deformations at condyle and the angle of the ramus regions higher than other areas (Fig.81).

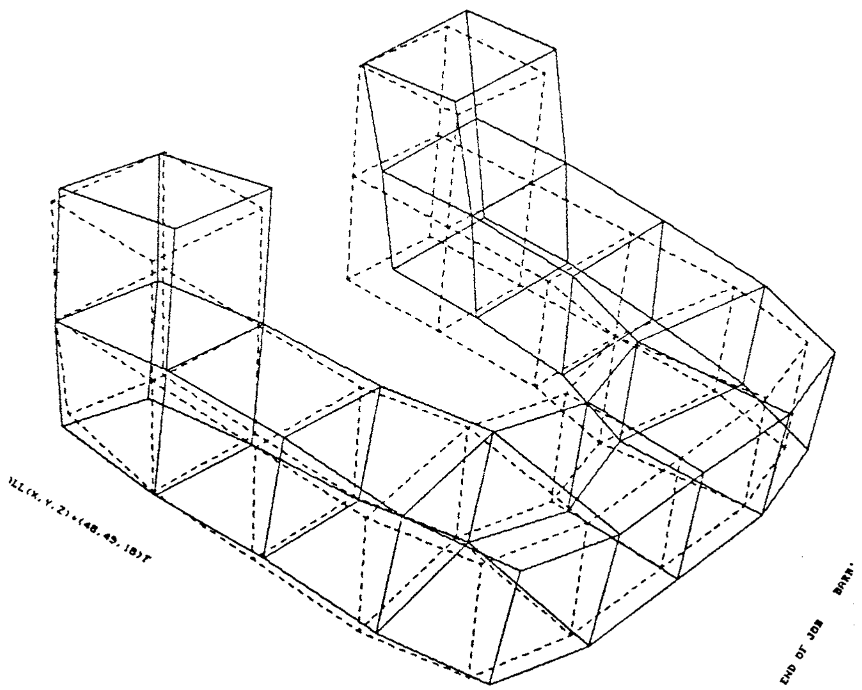


Figure 81 Preliminary primitive model, before and after deformation.

The model was restrained by three points at the base of the angle region and hence the distortion recorded could not have been realistically represented. As a general trend this primitive model has demonstrated that distortion was possible in structures similar to mandibular geometry.

After horizontal cross sectioning and measuring the mandible, first realistic model was created (Fig.82). The model was closer to anatomical model and applied forces were more realistic both in direction and magnitudinal values. Both "Metsap" and "MSS/Nastran" were utilized to analyze this structure.

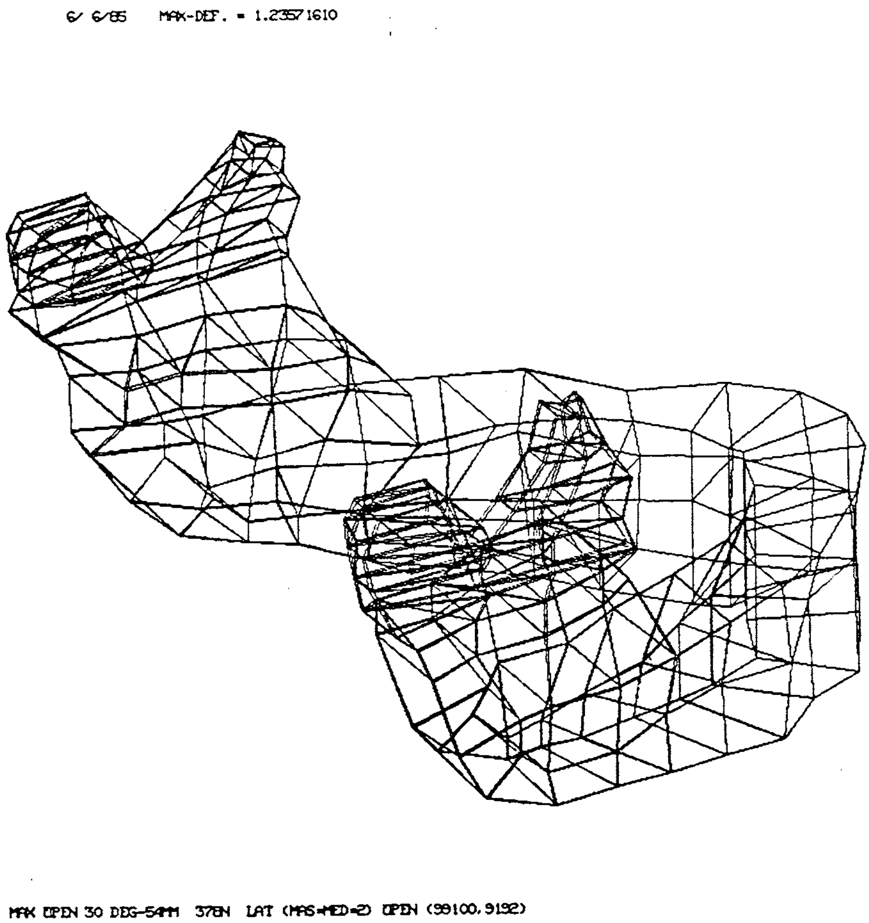


Figure 82 Three dimensional (deformed) model of the solid mandible (first generation).

The results showed that the deformation was of the order of 0.1 to 0.6 mm from first bicuspid to third molar region. The maximum distortion was observed in condylar region larger than 1.00 mm distortion for a material with Young's modulus of 10.3 GPa and Poissons ratio of 0.3.

Figure 83 shows the plot of distortion to the distance from the condyle centre. Differing values of Young's modulus were applied to analyze the extent of the deformation on the structure. As expected when cortical bone was considered to be the only phase in the structure, the model showed very low levels of distortion around the molar regions, for both in clench and open conditions. The opening produced slightly larger deformation than the clench condition.

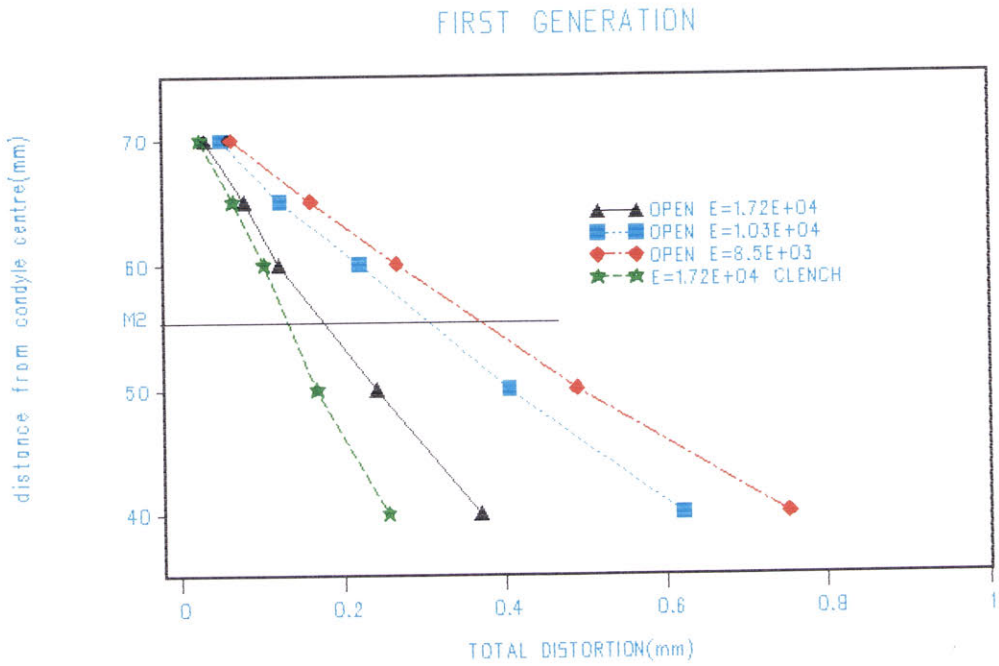


Figure 83 Distortion v.s the distance from condyle centre at various Young's modulus values under the clenching and opening conditions.

The third model was anatomically more accurate, because of the method used to digitize the cross sectioned mandible. Twenty node isoparametric elements also increased the accuracy of the finite element analysis. The undeformed and deformed structures are shown in Figure 84.

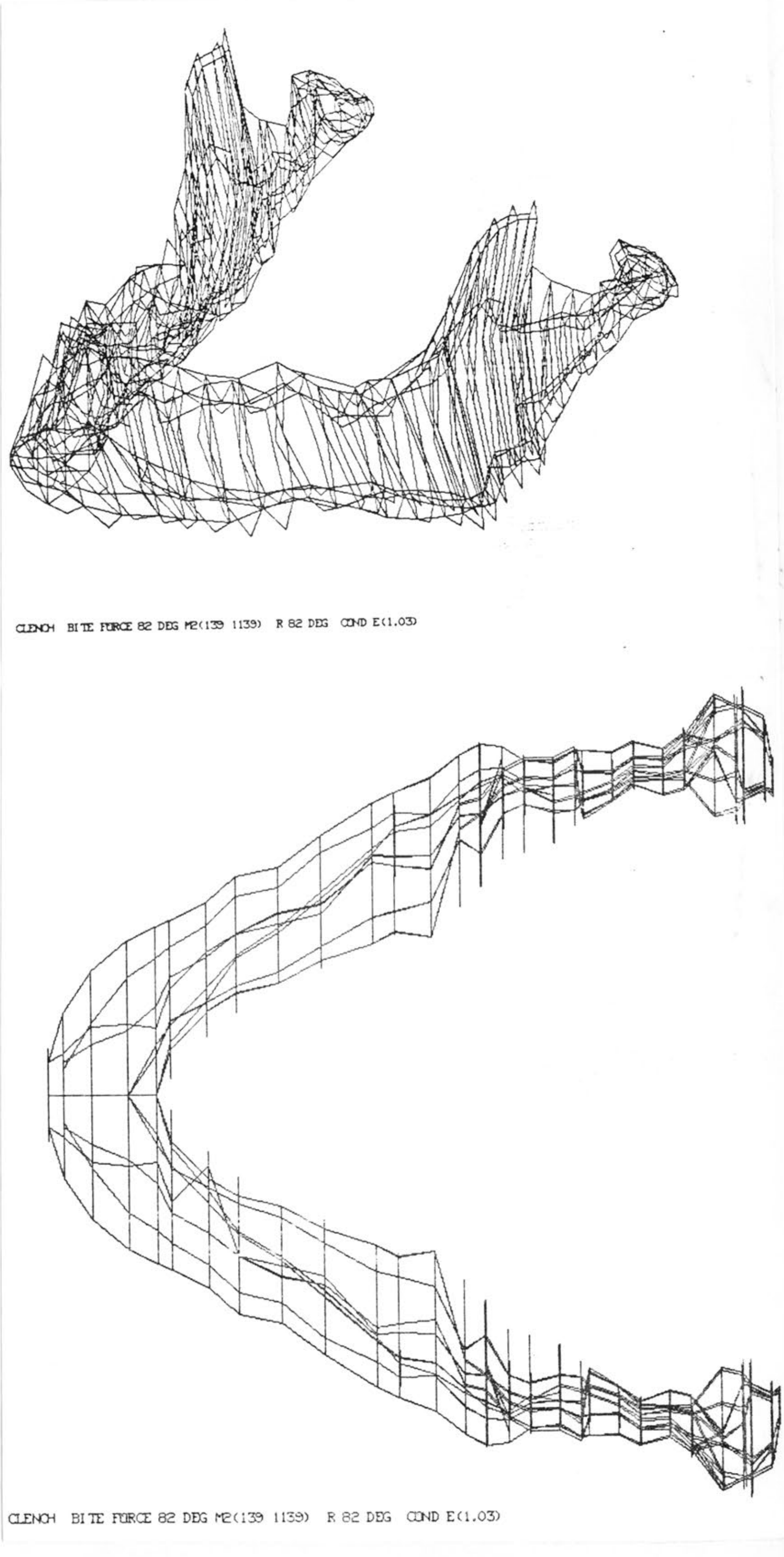


Figure 84 Undeformed (a) and deformed (b), 3-D model of the mandible.

The forces applied were similar to the previous model. When this composite model was analyzed using Young's modulus values of:

$E_1 = 1.72 \times 10^4$ MPa Cortical bone

$E_2 = 1.29 \times 10^3$ MPa Cancellous bone

wide opening of the mandible produced 0.35 mm distortion around the second molar region. Clenching demonstrated 0.15 mm distortion from second premolars down to the third molar region (Fig.85).

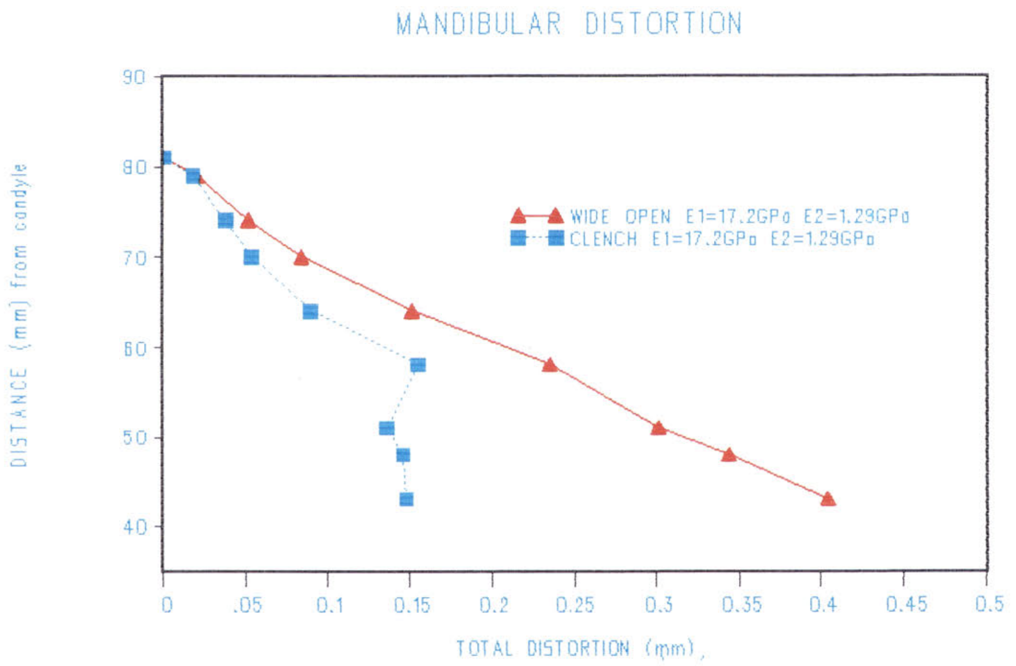


Figure 85 The distortional change of the mandible during wide opening and clenching (E_1 = 17.2 GPa, E_2 = 1.29 GPa).

To analyze the deformation, differing Young's Modulus values (Appendix IV) were utilized. Comparitive distortion values are given in Figure 86.

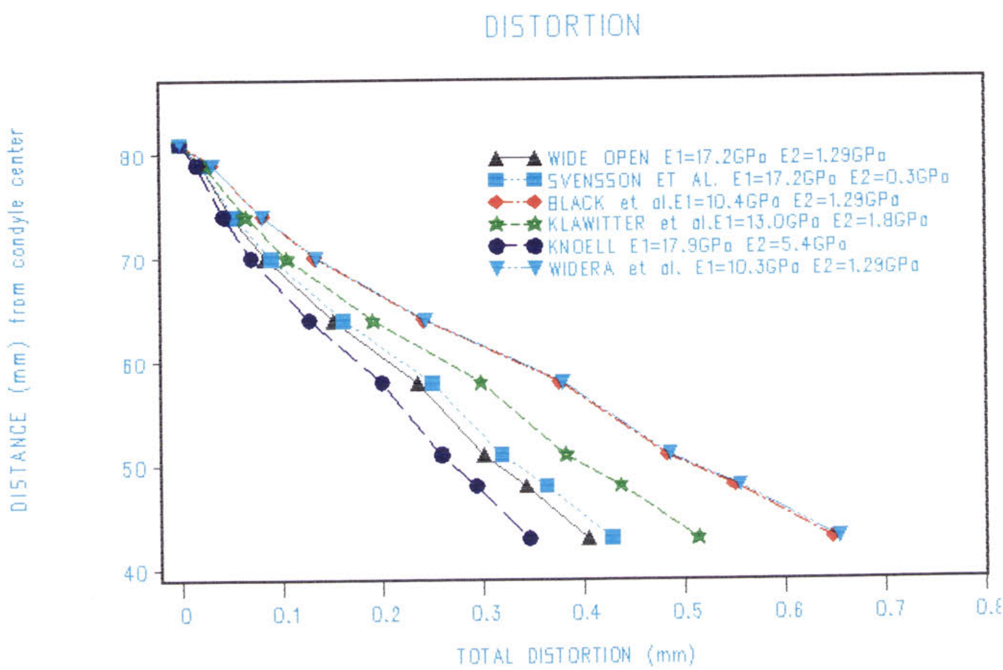


Figure 86 Differing Young's Modulus values applied to the distortion analysis of the mandible during wide opening.

As it can be seen from figure 86, the displacement values in second molar region (50 mm from the condyle center) varies from 0.25 mm to 0.5 mm depending on applied E values. The distortion data on clench with differing Young Modulus values are given in figure 87.

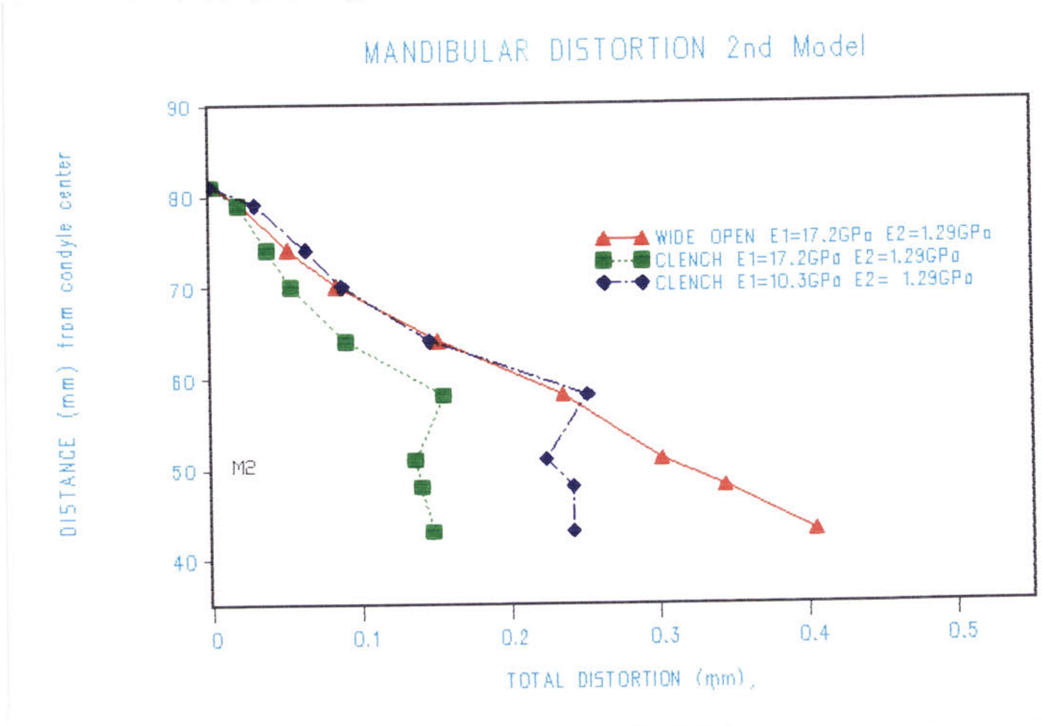


Figure 87 Comparative distortion values under differing Young's Modulus of elasticity during clenching, and during wide open.

As it can be seen from figure 87, during clenching the distortion becomes constant at 60 millimetres from the condyle centre which corresponds to the second premolar region. The total distortion of the mandible during clenching it is found to vary between 0.10 to 0.25 mm. It is very interesting to note the mean pressure change at the second premolar region (Fig.88). Comparison of the mean pressure during wide open and clench are given in figure 89 .

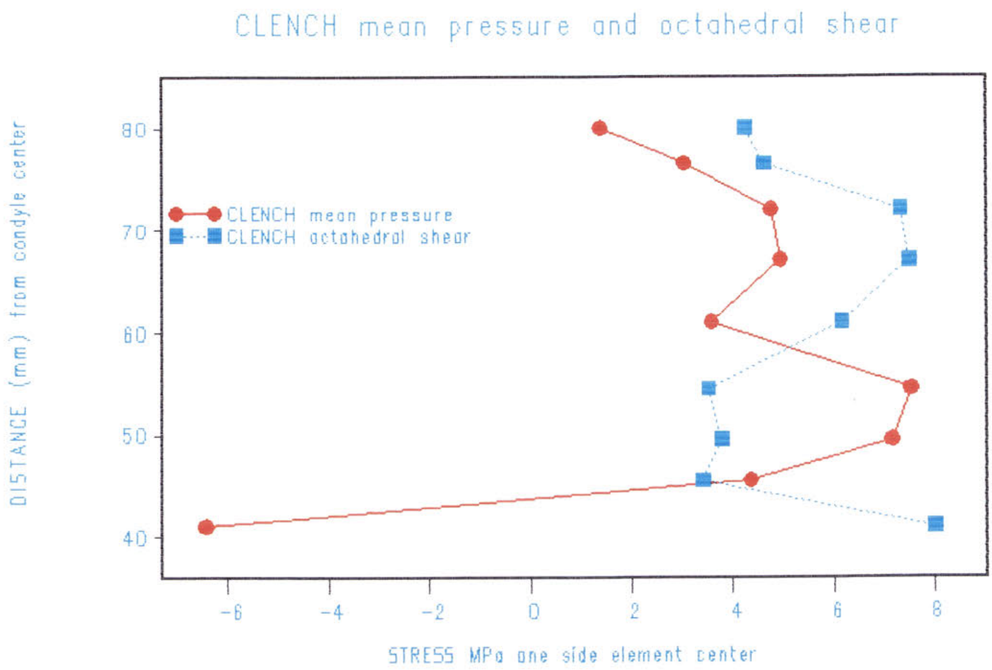


Figure 88 Comparison of mean pressure and octahedral shear during clenching along the most protruding point on bony ridge.

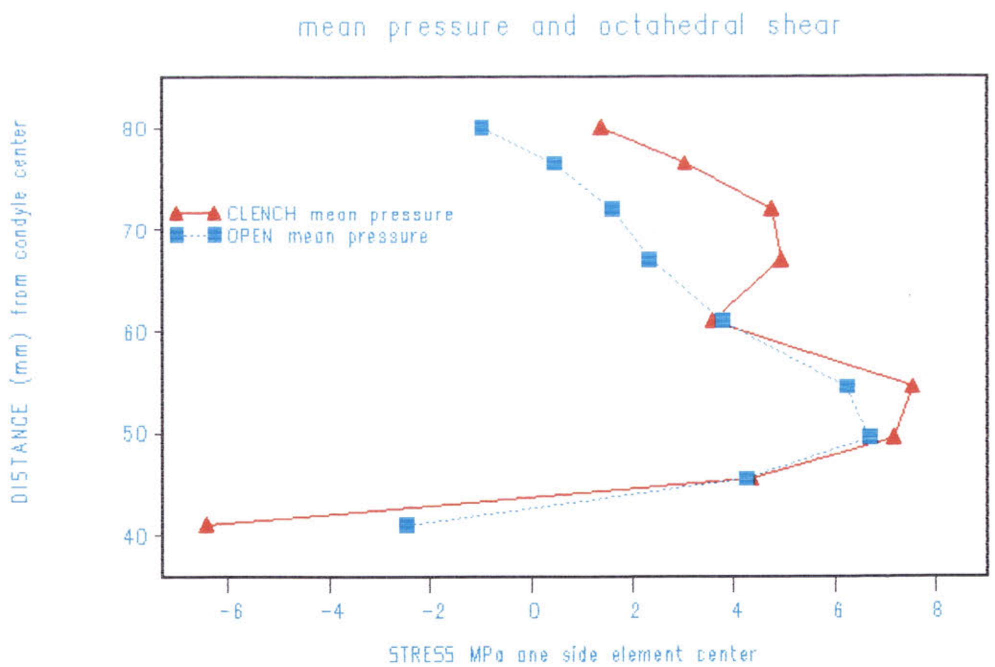


Figure 89 Comparison of mean pressure during wide open and clenching along the most protruding point on bony ridge.

Normal stresses and the shear stress along the alveolar ridge are given in figures 90 and 91.

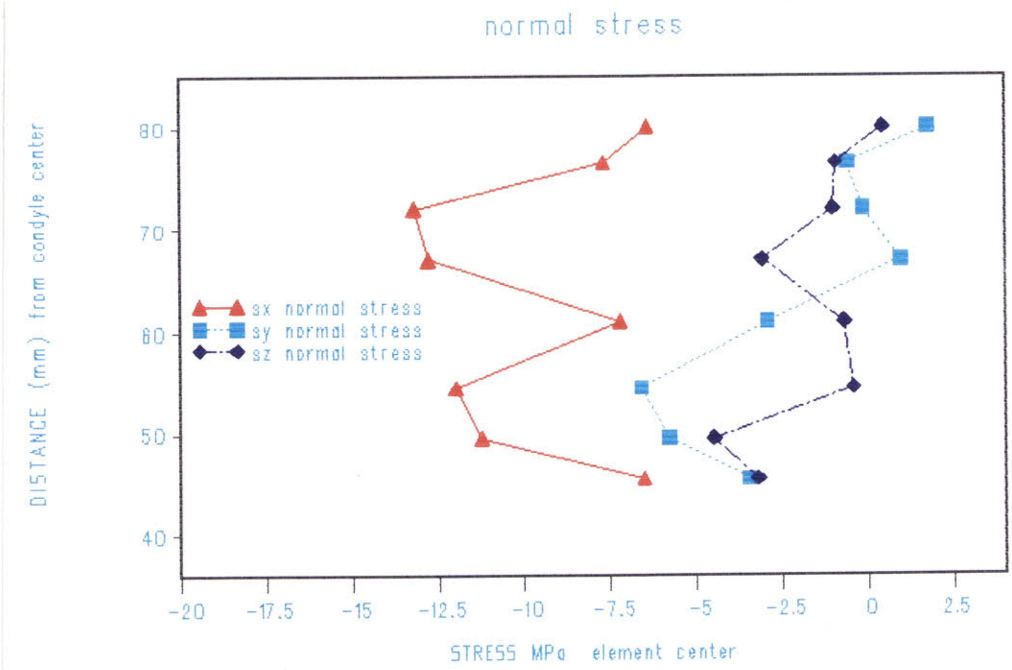


Figure 90 Normal stresses along the most protruding point on the bony ridge.

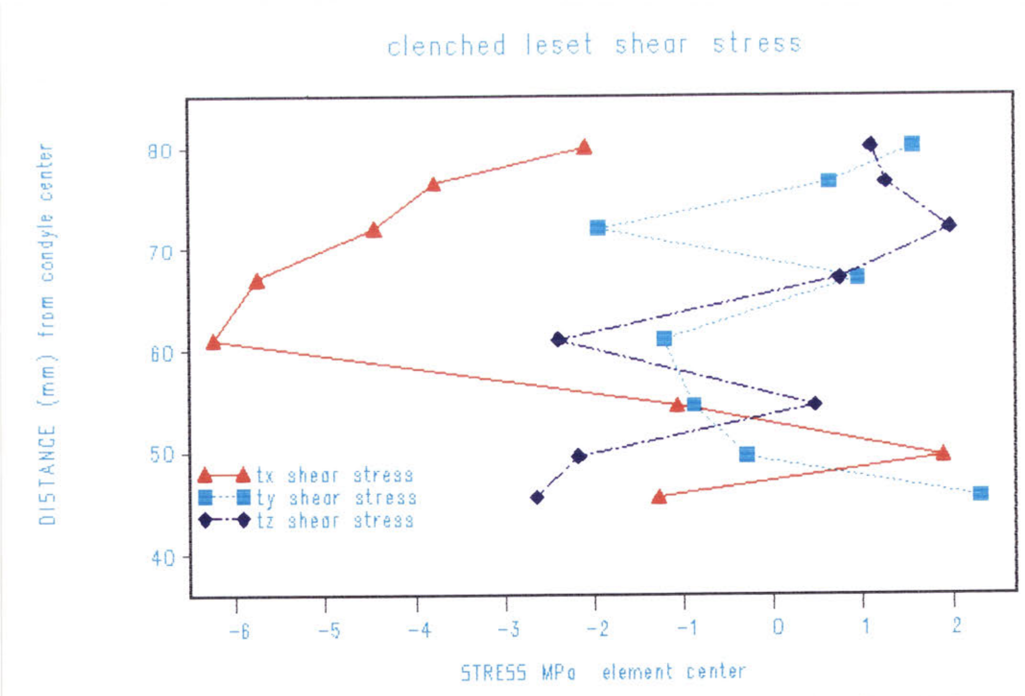


Figure 91 Shear stresses along the most protruding point on the bony ridge.

When the wide open mandible distortion values of the hollow cortical structured mandible (E^2 = 1.29 MPa), and solid cortical bone filled (E^2 = 10.3 GPa) mandible are compared distortion values of 0.35 to 0.56 mm can be observed at the second molar region. (Fig.92).

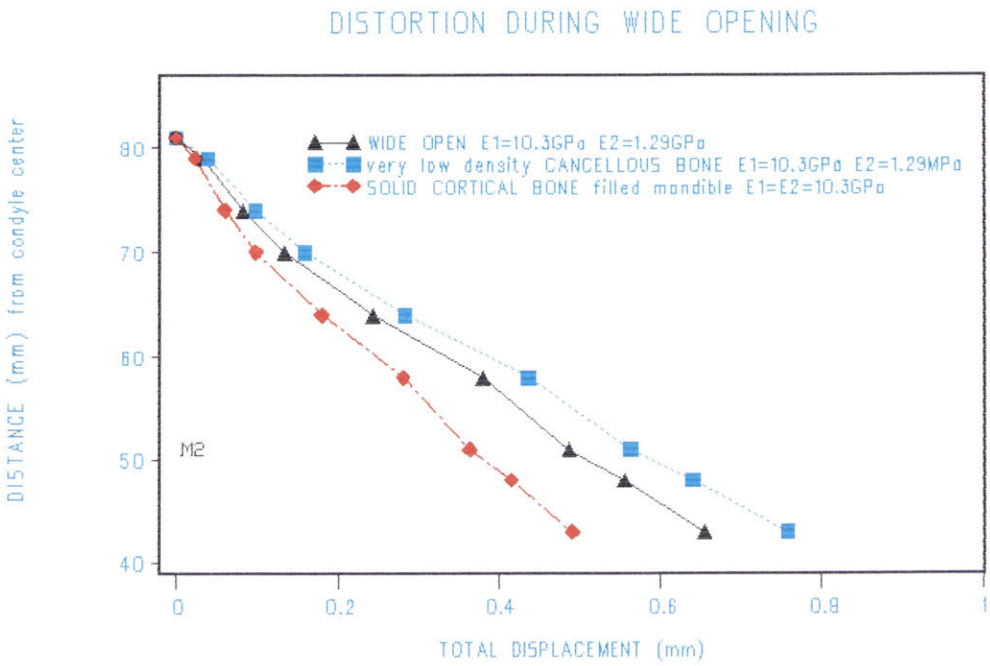


Figure 92 Comparison of hollow and solid mandibular bone, distortion values.

Comparison of the preliminary solid model with the improved model distortional values shows that, the use of overall Young's modulus of 8.5×10^3 MPa in the first model could give similar distortional values (Fig.93). This value as an approximation could be used in the further modelling of simple structures. Comparative distortional values of both models using this Young's Modulus value is given in Figure 94.

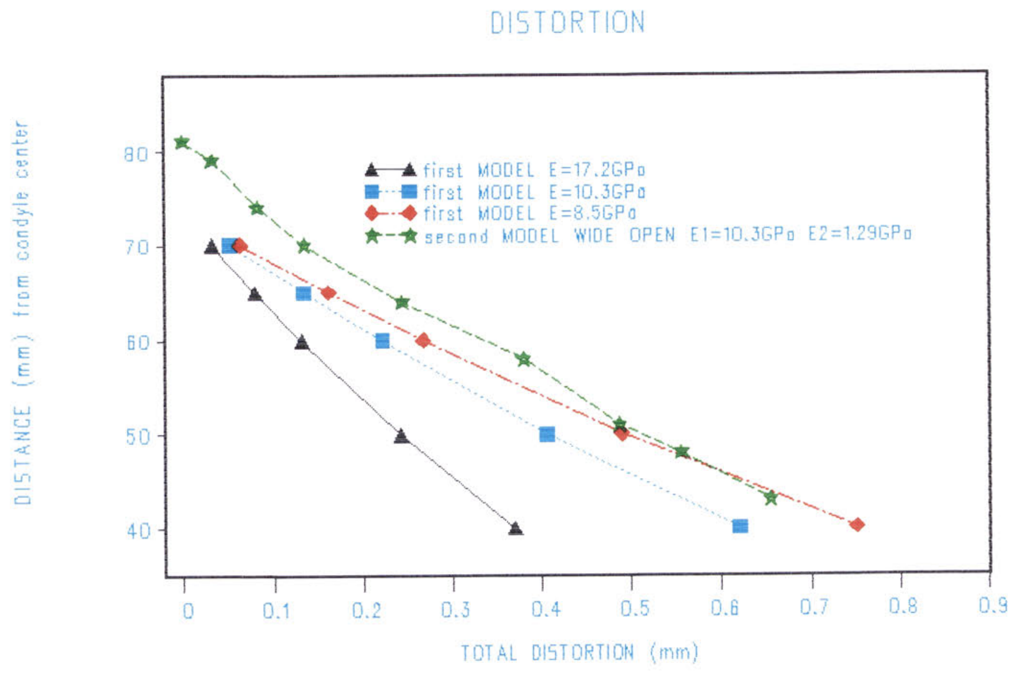


Figure 93 Comparative distortional values of previous solid model and the composite model at differing Young's moduli.

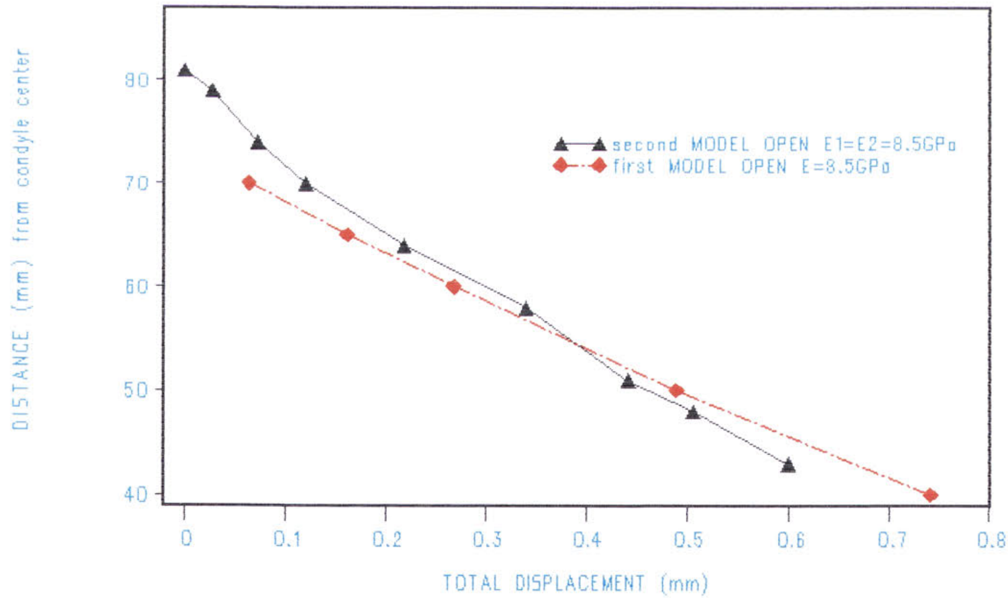


Figure 94 Comparative distortional values using the same Young's Modulus value for both models E(cortical) = E(cancellous) = 8.5 GPa.

When only the medial pterygoid muscle is assumed to be guiding the protrusive movement, larger mandibular distortions could be observed, compared to the open and clench conditions in the molar and bicuspid regions (Fig. 95).

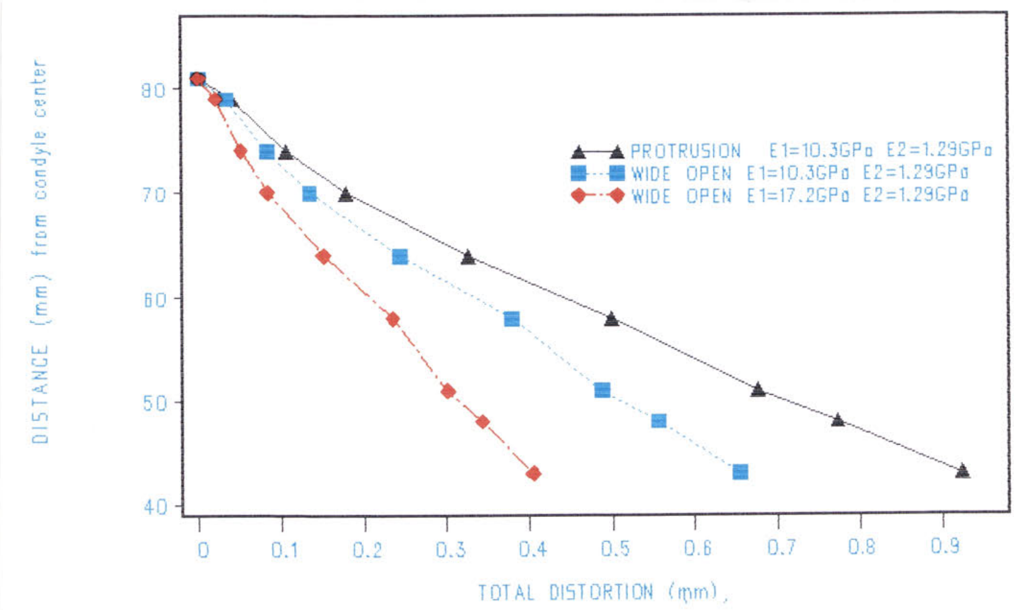


Figure 95 Comparison of distortion values of mandible during opening and protrusion.

When these protrusion distortion values are compared with the various published measured values (TABLE I), the present mathematical model well fits with the most of the measured values (Fig. 96, 97).

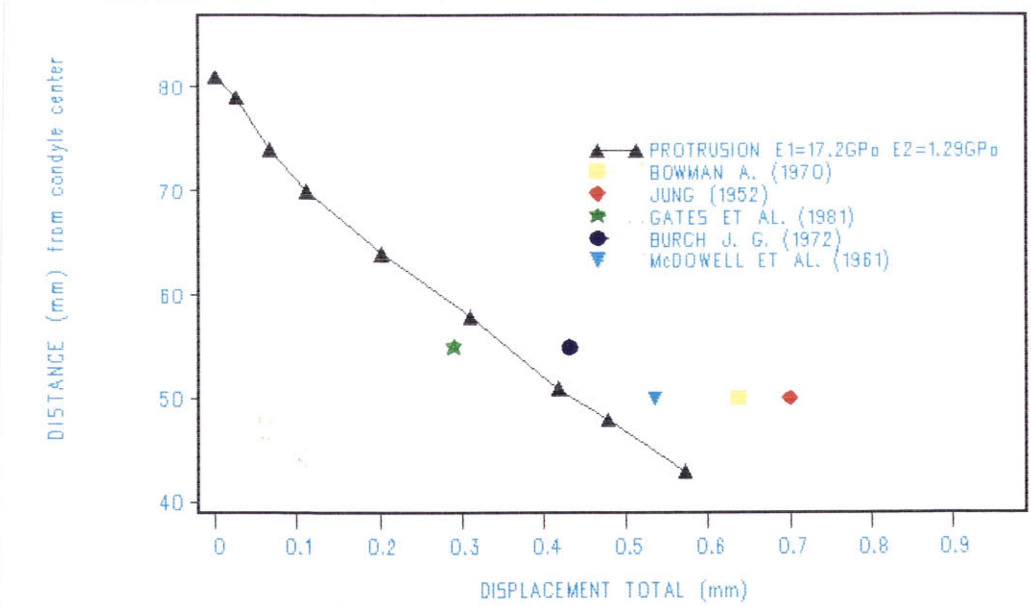


Figure 96 Comparison of the measured and the calculated distortional values during protrusion. E(cortical)= 17.2 GPa; E(cancellous)= 1.29 GPa.

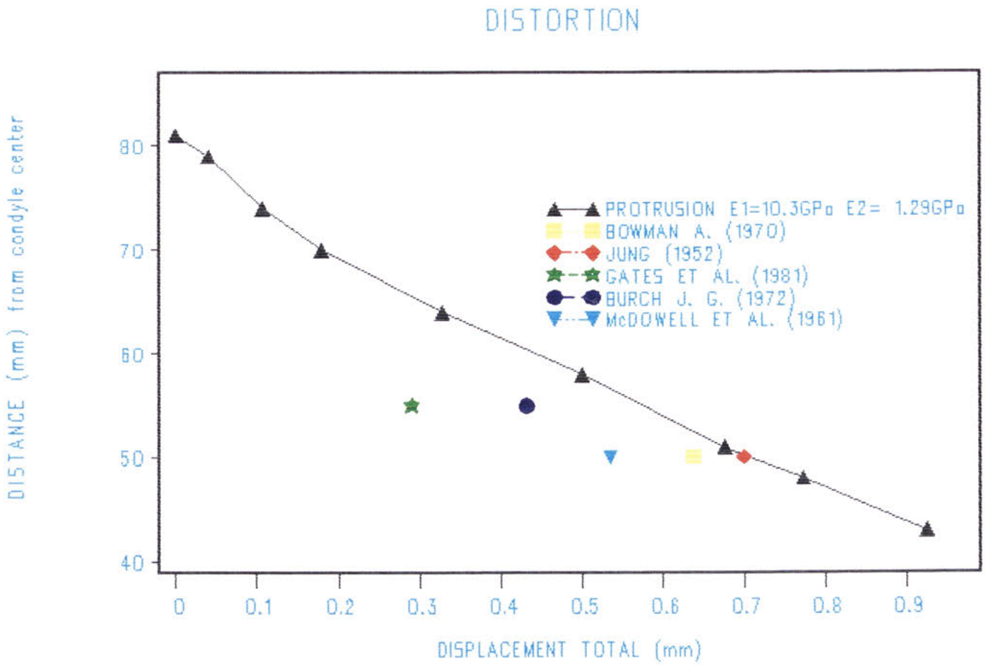


Figure 97 Comparison of the measured and the calculated distortional values during protrusion. E(cortical)= 10.3 GPa; E(cancellous)= 1.29 GPa.

Previously published measured values (TABLE I) for wide opening shows distortions of 0.078 to 1.0 mm around the second molar region. If the dental mandibular arch distortion measuring methods are compared, some of the very small amount of deformation can be attributed to very stiff splints or the inaccurate measuring methods used.

Siebert (1981) has measured the tooth movements during mandibular movements using electronic and pantographic techniques. He reported that elastic deformation of the mandible can give rise to three dimensional tooth movements of up to 60 μm in each direction.

It is possible that some of the workers were measuring only the tooth movement which were first noted by Muhlemann (1951, 1954), Zwirner (1949), Picton (1957) and Parfitt (1960).

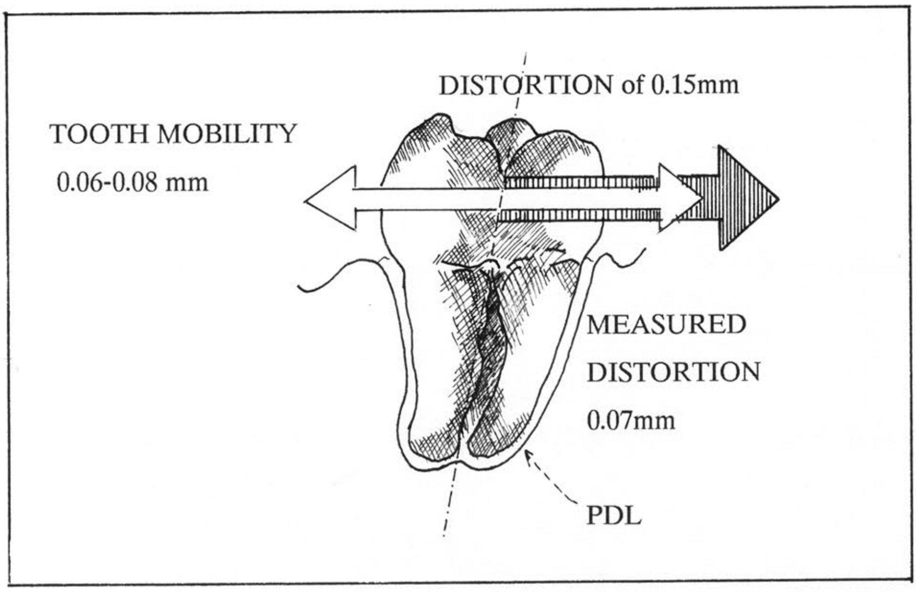


Figure 98 Tooth movement and distortion (one side).

It is even possible that some of the workers might be substracting the opposite tooth movement and measuring only the net distortion (Fig.98). Bowman's (1970) work is noteworthy on overcoming this problem.

If we do not take into account these low measurement values and extremely large values, average of 0.227 mm distortion for the first molar and 0.482 mm for the second molar can be calculated for the wide opening movements.

If Fishman's (1976) value is also omitted from the average calculations- because the mandibular distortional measurements were carried out between the mandibular second right and left first molars- then the average distortion in the second molar could be shown to be 0.356 mm (TABLE VIII).

TABLE VIII

Comparative values of the distortions (mm) at first and second molar regions, both as reported and corrected during WIDE OPENING.

	FIRST MOLAR			SECOND MOLAR	
REF*	REPORTED	CORRECTED	REF*	REPORTED	CORRECTED
(6)	0.438	0.438	(1)	0.35	0.35
(8)	1.00	-	(3)	0.421	0.421
(9)	0.224	0.224	(4)	0.07	-
(10)	0.076	-	(5)	0.09	-
(11)	0.78	-	(7)	0.297	0.297
(13)	0.105	0.105	(12)	0.861	-
(14)	0.093	-			
(15)	0.142	0.142			
AVERAGE	0.397	0.227		0.3094	0.356
THIS **	0.265			0.32	
WORK ***	0.425			0.51	

TABLE IX

Comparative values of the distortions (mm) at first and second molar regions, both as reported and corrected during PROTRUSION.

	FIRST MOLAR			SECOND MOLAR	
REF*	REPORTED	CORRECTED	REF*	REPORTED	CORRECTED
(6)	0.61	0.61	(1)	0.70	0.7
(9)	0.432	0.432	(2)	0.50	0.5
(15)	0.29	-	(3)	0.535	0.535
	-	-	(4)	0.09	-
	-	-	(7)	0.637	0.637
AVERAGE	0.444	0.521		0.492	0.593
THIS **	0.36			0.432	
WORK ***	0.575			0.71	

* Full references are given in TABLE I (page: 8).

** E(cortical)= 17.2 GPa; E(cancellous)= 1.29 GPa.

*** E(cortical)= 10.3 GPa; E(cancellous)= 1.29 GPa.

During protrusion average value of 0.521 mm in the first molar region and 0.593 mm distortion in the second molar region can be postulated.

When previously published values of distortion during opening are compared (TABLE VIII) (Fig.99, 100) the total displacement can be shown to be between 0.15 to 0.6 mm around the second molar region.

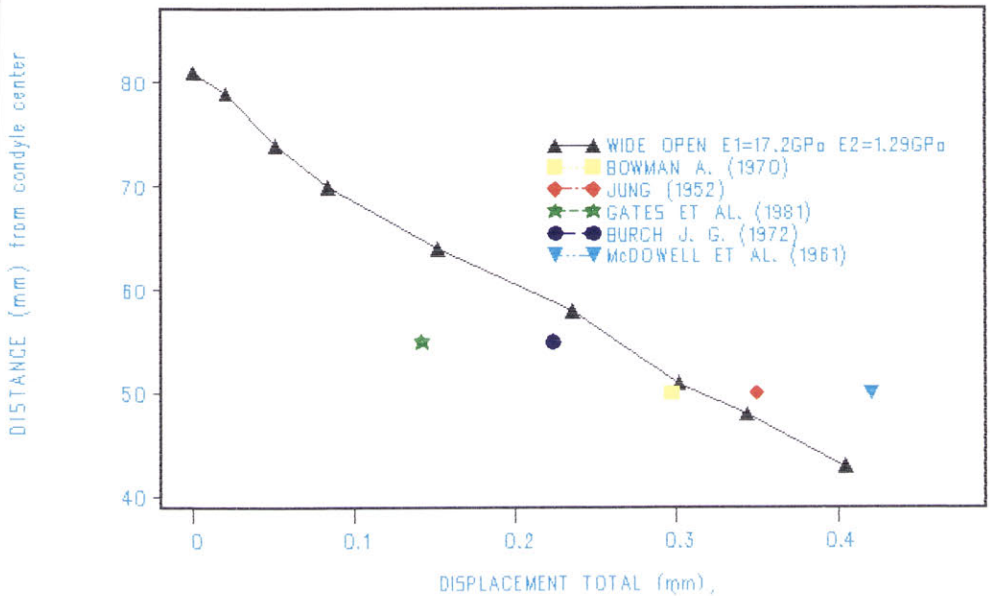


Figure 99 Various measurement values of distortion during wide opening movements. E(cortical)= 17.2 GPa; E(cancellous)= 1.29 GPa.

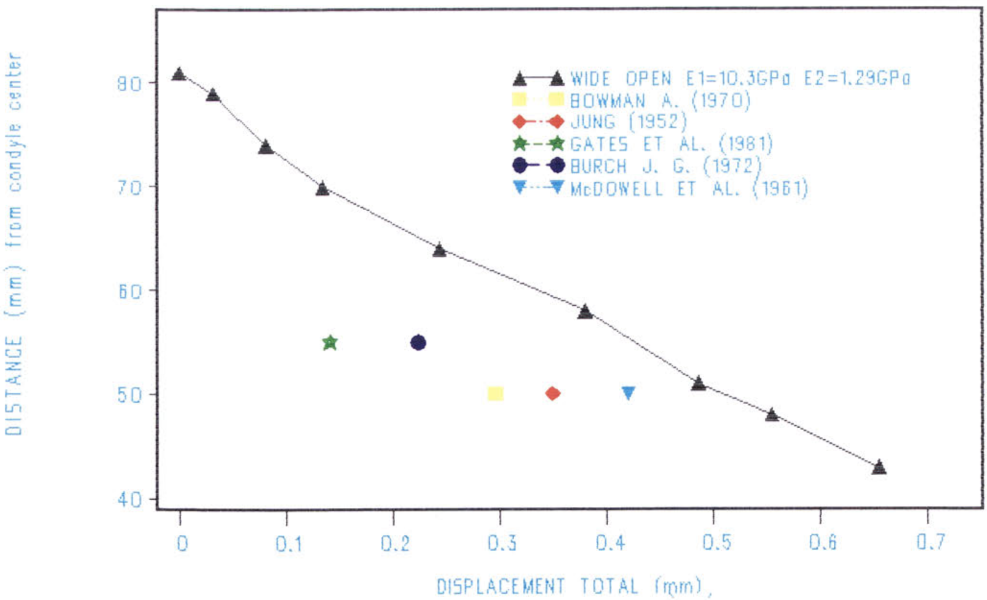


Figure 100 Various measurement values of distortion during wide opening movement. E(cortical)= 10.3 GPa; E(cancellous)= 1.29 GPa.

CHAPTER 11

CONCLUSIONS

The biomechanical aspects of the functional distortion of the mandible have been studied in the present thesis. Three dimensional modelling of the mandible has been attempted using computer aided design and digitizing methods.

The three dimensional modelling method by using CAD, digitizers and finite element pre processors could be utilised in further analysis of the mandible and other biomechanical systems.

Based on the electromyographic and anatomical observation data, forces acting on the mandible and its various movements (clench, wide open and protrusion) were analyzed through a mathematical model. To date this is the most realistic and advanced mandible model. Distortions in the mandibular bone during various movements were studied using a finite element model in which three dimensional solid elements were employed for both cortical and cancellous bone.

The distortion of the mandible by using two differing models has been found to occur and the amount of the distortion is dependant on the thickness of the cortical bone, geometry and the size of the mandible. If various Young's modulus values are applied to the structure, or the design of the condyle is modified, different values of distortion could be observed. These results indicate that differing values of mandibular distortion could be observed from a child to an adult or between male or female species.

Stress analysis of the deformation of the three dimensional model shows, displacement values of (0.265-0.425mm) in first molars and (0.32-0.51mm) second molars during wide opening movement and (0.36-0.575mm) in first molars and (0.432-0.71mm) in the second molars region for protrusion.

When published values of distortion are compared with this work, the calculated displacement values are in very good agreement with the previously measured values.

This good correlation with the published values might indicate that the assumed muscle force magnitude and directions and Young's modulus values are satisfactory and could be reutilized in future mandible models until the appropriate detailed data are published.

The range of mandibular movement and distortion during opening and protrusion could present some important clinical problems:

1. The amount of mandibular distortion during opening and protrusion is sufficient to effect the fit of a removable prosthesis and can put cyclic internal stresses on abutment teeth of a bilateraly fixed prosthesis.

2. A dental impression taken at a wide open position will not be correct base for a dental prosthesis in most intercuspal position.

3. Bilaterally fixed bridgework, fabricated on a cast made from an open mouth impression, could present as occlusal interferences when the patient applied muscle force during mastication.

4. Fixed prosthesis involving several teeth and implants would not completely inhibit mandibular deflection. Inhibition of mandibular flexure could increase as more teeth are splinted and more rigid attachments are used, however there would be a detrimental effect to the abutment.

Some Temperomandibular joint problems and disorders have been proposed to be generated from muscular or occlusal disorders, these or other related medical problems could be related to excessive mandibular distortions and should be further investigated by using the developed mandibular model.

In the last twenty years many different types of implants have been used in prosthodontics for the replacement of missing teeth (tooth replica implants, subperiosteal, endosseous, endodontic-endosseous implants).

In endosseous implants it has been shown that the forces applied to the functioning implants can result in relative motion which causes fibrous encapsulation. In an extensive study of vitrous carbon tooth-root endosseous implants Schnitman and Shulman (1980) have found a high incidence of implant fracture, gingival recession and exfoliation.

There is some controversy over the question of the success of diodontic (endodontic-endosseous) implants. Histologic studies in humans have shown that (Simon and Frank, 1980) endodontic implants can become encapsulated by fibrous connective tissue. It is thought that implant loading due to exceeding physiologic loading results in the progressive thickening of this fibrous tissue capsule leading eventually to implant instability and failure (Maniatopoulos et al.,1986).

However according to another group of investigators, in a properly inserted diodontic implant, the implant could place the fulcrum of movement deeper, thus stabilizing the tooth. It is hermetically enclosed within the bone and the root of the tooth in a sterile environment and there is no contact with oral fluids or bacteria (Vajda, 1987).

The success ratio of implants in dental and maxillofacial surgery is somehow lower than other implant applications. The reason for this might be the difficulty of achieving the desirable response from the adjacent tissues. Tissue reaction in this situation is influenced by factors such as the transgingival location, infection, functional loading, inadequate bone support and insertion techniques and implant design. Oral implants probably encounter the more adverse conditions any surgical implant has to face and most of the problems encountered singularly in other situations could present themselves together, providing a relatively poor prognosis.

A more precise understanding of the interplay between bone mechanics and the structural properties of bone in function is necessary. Actual biomechanical stresses transmitted to the tissues, the acceptable range of stresses in the tissues, the magnitude of the relative motion between the implant and the tissues and the determining factors of the gingival interface response must be investigated and related to implant success.

In order to develop biomechanical criteria for implant design, a comparison between stress distribution experiments on the mandible invivo should be carried out using the present 3-D model. This model could be used to assess the efficacy of an implant, reasons for its possible loosening and provide input for improved dental prosthetic designs and implant techniques.

At present the design criteria for implants is based on "rigid fixation" and the materials chosen are usually stronger and stiffer than the bone.

Present investigation has shown that mandibular bone flexes. During various mandibular movements, oral implants if not properly designed and inserted could be subjected to an alternating load oscillating through zero. The results are micromovements between the implant and the bone, which cause an increasing loosening of the implant. This condition and subsequent fibrosis, might facilitate infection and its failure.

A new design criterion should be developed on the basis that, under certain circumstances it is the deflection (EI) and flexure of the bone and the adaptability of the implant to these movements is important rather than only stiffness and strength.

Implants should be designed and used under two different considerations, according to their need and suitability. Some prosthesis should induce "total immobilization" for bone regeneration and rapid recovery, and some should conform to the physiological requirements by matching the deformation. In addition to design modifications in prosthetic devices, new biomaterials should be developed to accomodate this need.

CHAPTER 12

SUGGESTED FUTURE CLINICAL CONSIDERATIONS

12.1 Clinical

The clinical implications of mandibular distortion and arch width changes associated with muscle activity are significant when occlusal surfaces are to be restored. Clinical and laboratory procedures should take this into consideration.

A closed-mouth impression technique and non muscle-RAP recording technique could be suggested. Alternatively, individual castings should be indexed in the mouth with minimal opening of the mouth (20 mm), and new casts remounted within the articulator. Cuspal interferences could then be eliminated.

Dental clinicians have experienced problems with retention and stability of full arch mandibular prosthesis. The fact that the mandible bends during various movements, presents one possible explanation as to why a prosthesis may come loose. A fixed mandibular prosthesis may include a stress breaker to allow independent lateral expansion and contraction of the mandible.

Similar attitude to the intra mandibular implant devices should be applied during the design stages. Accomodation of the mandibular flexure should be one of the most important design criteria, during the development of this type new clinical devices.

Some later stage temporomandibular joint dysfunction might arise from the stresses acting, with the use of corrective devices especially in young children. Detailed objective research and careful statistical analysis needed to determine the long term effects of these orthodontic corrective devices in young children.

Some of the previously mentioned functional disturbances or dysfunction syndromes of the masticatory system can be due to the excessive functional distortion of the mandible. This flexion and displacement might play an important role on some of the mandibular or temporomandibular joint dysfunctions. An appreciation of this fact might further improve the understanding of these types of disorders and might help to select an effective treatment.

12.2 Further Research

Several improvements on the developed three dimensional model could be suggested to provide better insight in the fields of, biomechanics, oral implantology, periodontology, prosthetic dentistry, maxillofacial reconstructive surgery, temporomandibular joint dysfunction problems and periodontic corrective device methodology.

i) The model could be improved by using nonhomogeneous, nonlinear material properties for both cortical and cancellous bones. Existing data on femur can be utilized or further analysis on the mandibular bone density and architecture could supply the approximate relevant data.

ii) A mathematical model of the tooth, periodontal membrane and surrounding structures can be further constructed on the existing model. All relevant mechanical properties should be systematically added into the calculations.

From these mathematical models, theoretical stress levels and distortions can be calculated. All such calculations, whenever possible, should be compared with published data or should be verified by clinical or animal experimentation.

iii) At the present time it is not possible to place strain gauges in the periodontal membrane to measure stress-strain distribution. If the stress level could be accurately determined in different areas of the periodontal membrane, it may offer the best opportunity for correlating force application on a tooth with tooth response. At clinical level, phenomena such as rates of tooth movement, pain response, tooth mobility, alveolar bone loss and root resorption may be studied.

iv) Mandibular model dimensions could be modified to represent, child, female and male. Distortions and stresses can then be calculated and compared.

v) Some teeth could be omitted from the completed model, to calculate the effect of the tooth removal on the distortion and possibly on bone remodelling.

vi) Muscle insertions in the model can be critically reviewed and the muscle origins and insertions could be distributed to an area representing more natural force application.

vii) The large number of variables acting during the production of a bite force might be analysed by simulating differing parts of the muscles simultaneously, which could be imposible to observe during experimental measurements.

viii) Muscle forces during protrusion should be modified to include temporomandibular ligament and opener force should be increased to produce more realistic protrusive movement.

ix) Addition of prosthetic devices into or onto the mandible model and analyzing the related mandibular deformation could help to design new types of dental devices.

x) Extreme atrophy of the mandible of edentulous patients still presents a hard challenge to oral surgeons. For subperiosteal implants, as for the treatment of mandibular fractures by internal fixation, the three basic principles remains the same (Spiessl, 1974).

a) Absolute stability
b) Ideal adaptation
c) Optimal vascularization.

During recent years, progress has been made in the development of certain ancillary techniques and operative procedures.

Compound mandibular fractures may be combined with defects involving the alveolar process and/or the body of the mandible. Presently to create a state of stability from the interplay of forces, a mechanism which counteract and neutralize all the forces involved is usually constructed. This mechanism is a combination of tension band and stabilization plate. These are called DCP (dynamic compression plate) and EDCP (excentric dynamic compression plate). In addition to plates some oral surgeons also use wire ligatures and wires (Schmoker and Spiessl, 1973).

The design and modifications of these methods and materials takes extensive amounts of experimentation and time.

The use of mandibular model can include these fixation plates or wires and the stresses can be calculated without expensive and painful experimentation and the selection of the design and materials could be simplified.

xi) The finite element model can be utilized for further modelling and analysis of the mandible or other biomechanical systems.

At present there are three promising methods which could be incorporated into the biomechanical analysis.

a) Leitz three co-ordinate CNC measuring machines, which continuously scan for precision measurements could be utilized without cross sectioning the bones. Data collected could be screened and fed into, the input data files which could generate the FEA nodal points.

b) High performance digital imaging cameras which convert stationary pictures and objects into digital form via a linear sensor array has been marketed by Kodak company which could be utilized for nodal point generation. With the further advances in the robotics industry the number of equipments with similar or better abilities could increase quite effectively in the near future.

c) Surface imaging from computerized tomographic scans (CAT) has been around from 1976. Computer tomography is quite an accurate technique to determine the geometry of bones. For finite element modelling, this data could be screened and then fed to FE pre-processors.

d) Also the existing model could be simplified and fed into FEA capable Personal Computers that could be utilized by the dentists, dental labs, hospitals and dental surgeons.

Some further work strongly recommended.

This thesis has explored the possibility of applying theoretical mechanics and finite element analysis, to the study of mandibular distortions. In the past, the physical and histological sciences which have made the most rapid and rewarding advances have been those that have been able to quantify their observations.

Although this investigation in the biomechanics of the mandibular distortion has great promise for further work, there are certain dangers in employing mathematical methods to describe biologic phenomena. Particularly, mathematical over simplification of highly dynamic and variable vital structures and reactions can mislead as well as inform.

It is therefore necessary to compare biomechanical assumptions with observations on both clinical and histologic levels. A multidisciplinary approach of this type offers the best hope for solving oral and dental problems related force systems and mandibular distortion.

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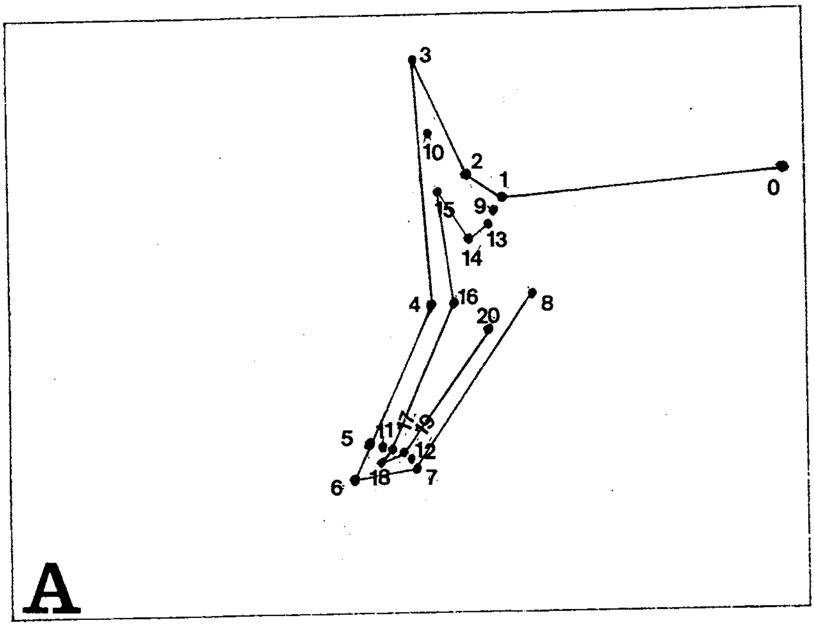
APPENDIX I

Digitizing method and finite element nodal point generation. Following figures represents, the final twenty nine plots of the digitized cross sections.



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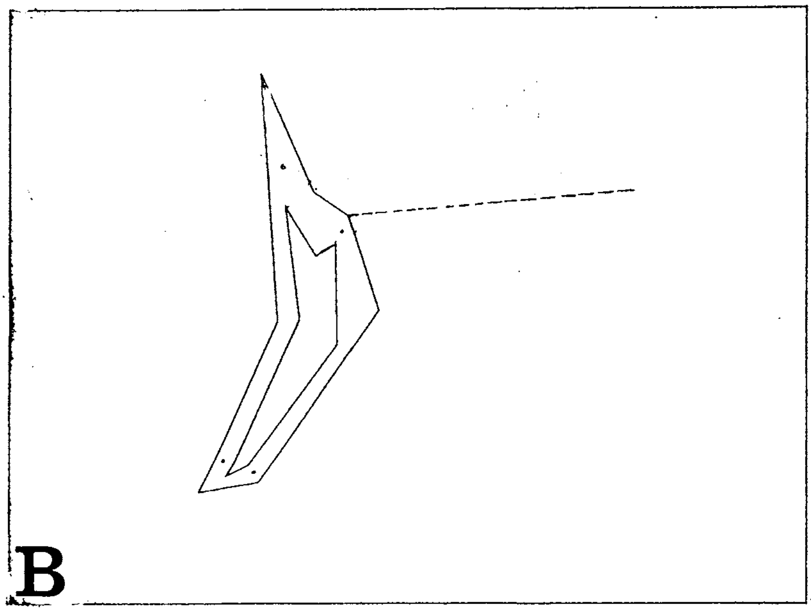

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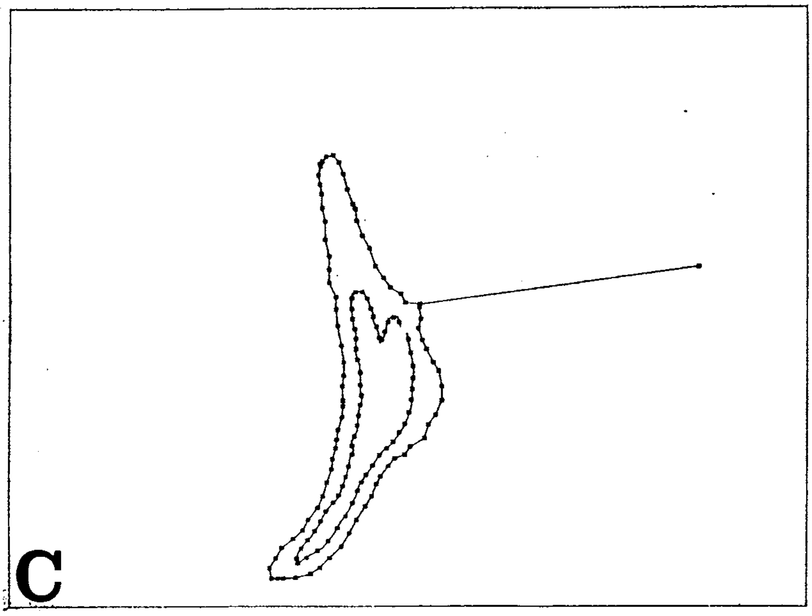

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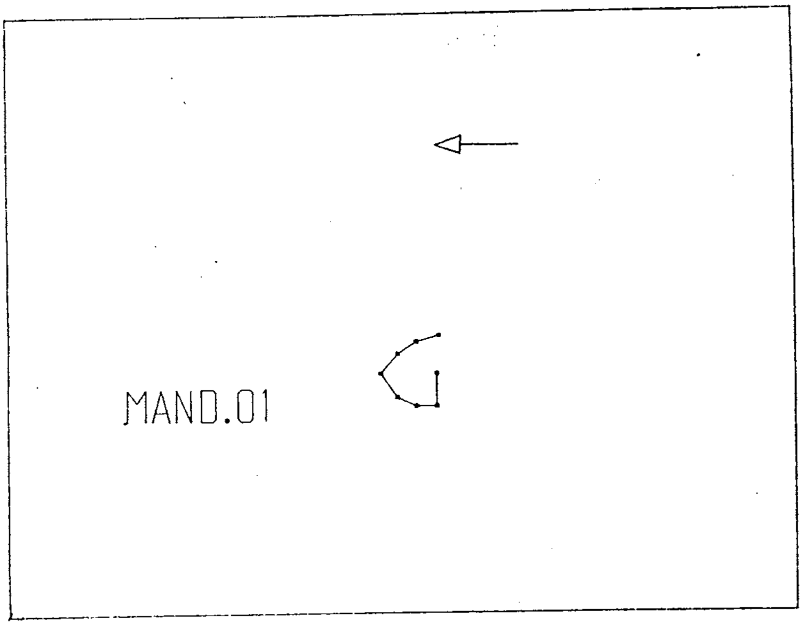
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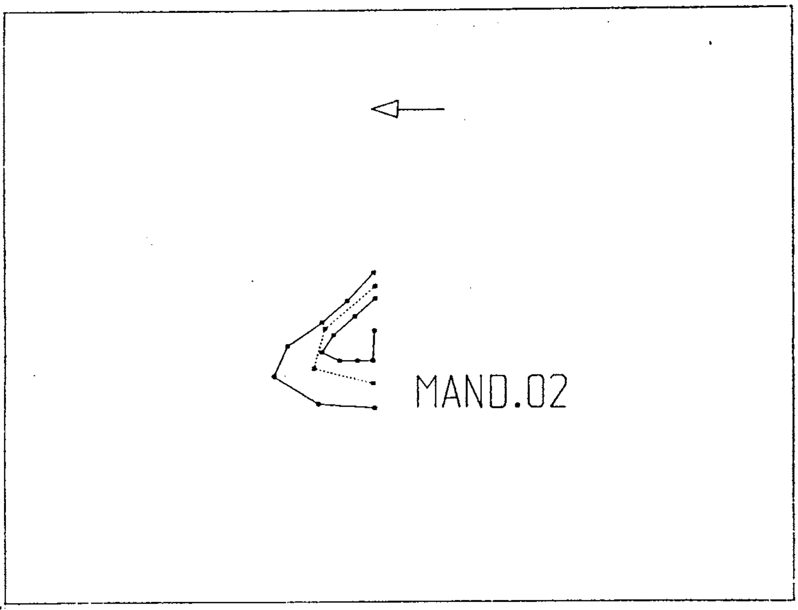
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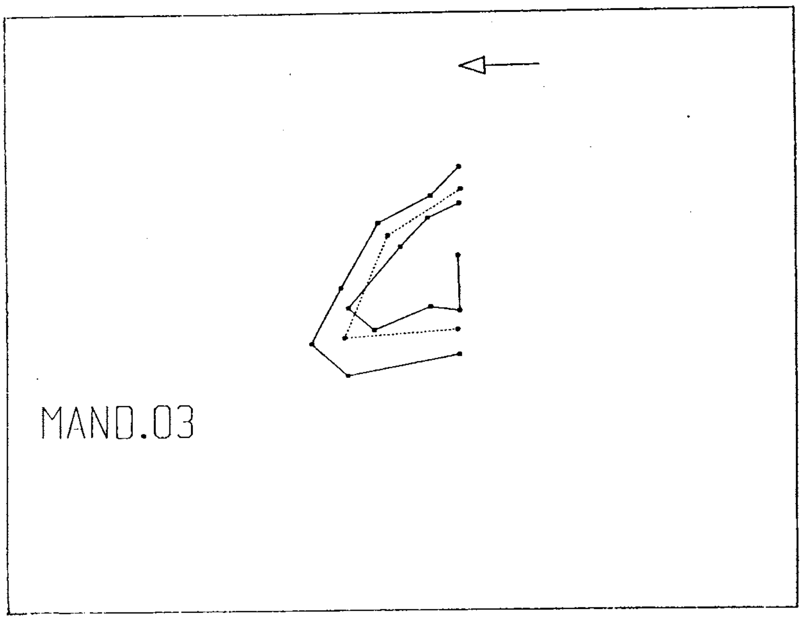
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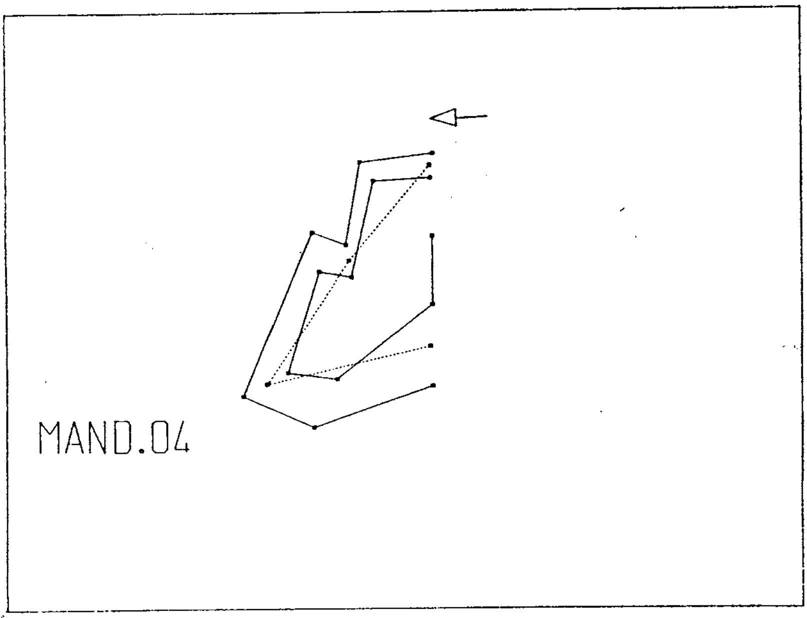
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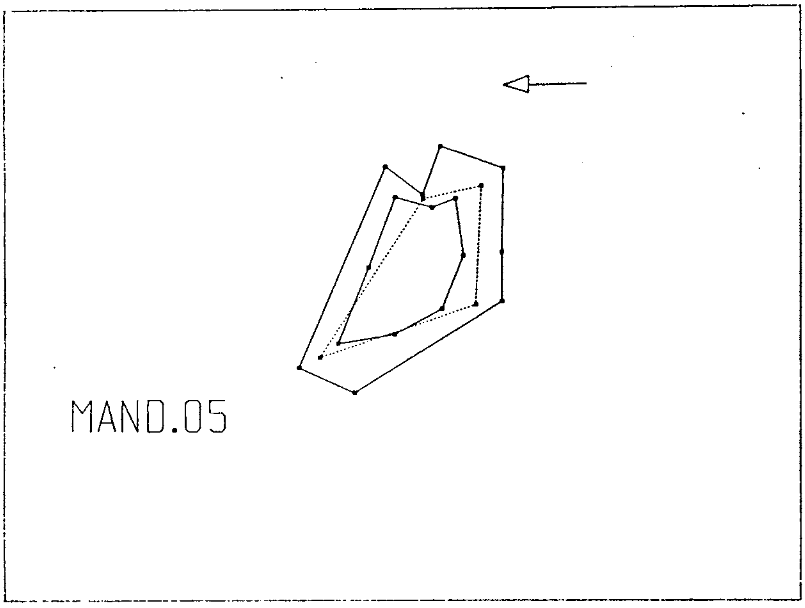
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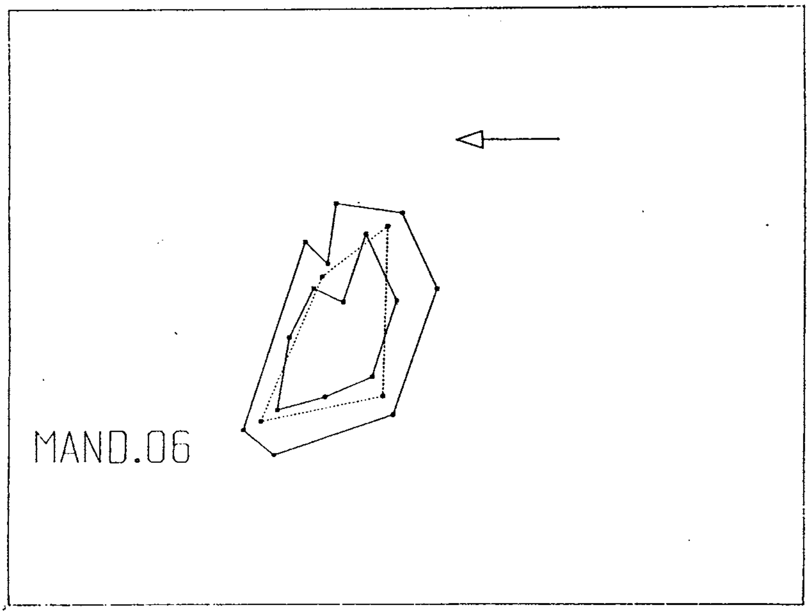
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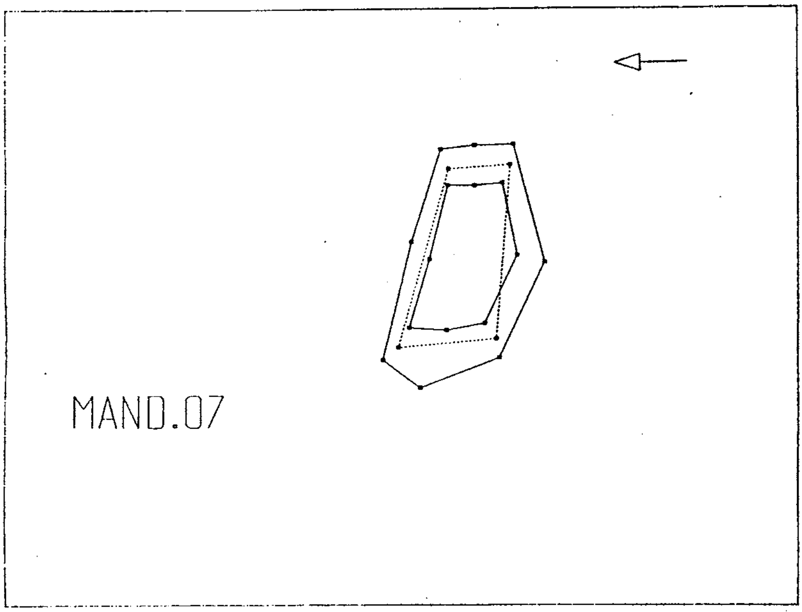
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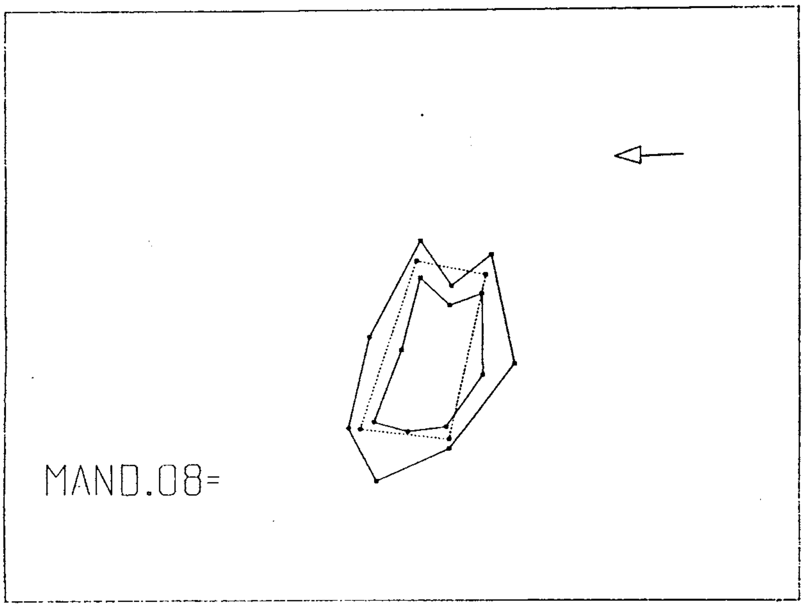
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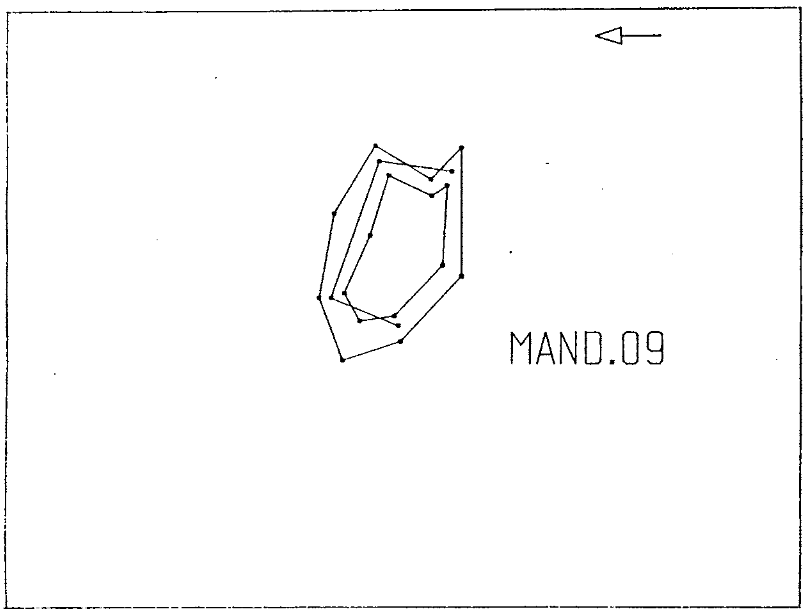
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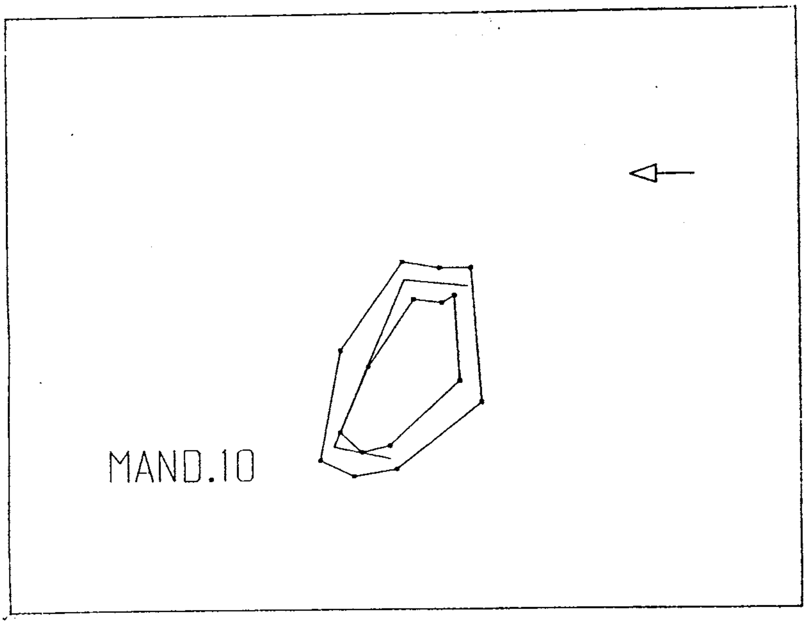
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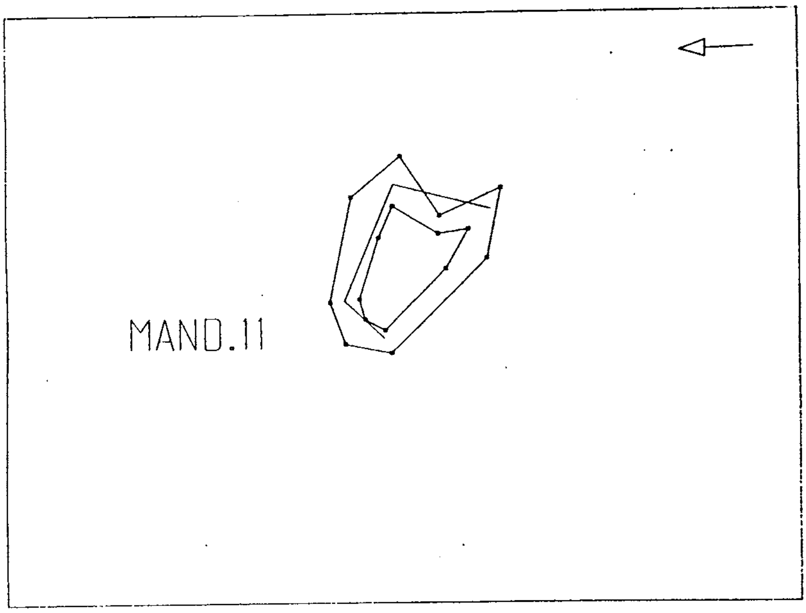
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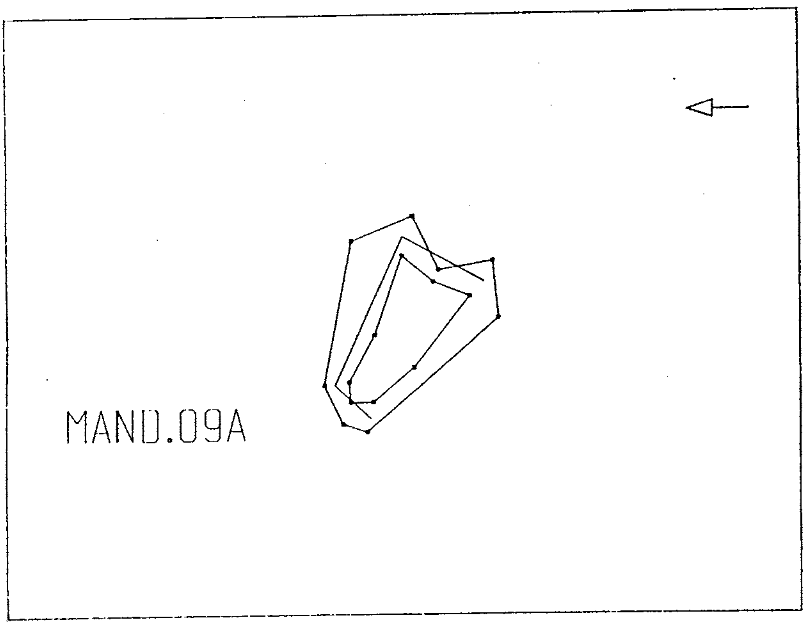
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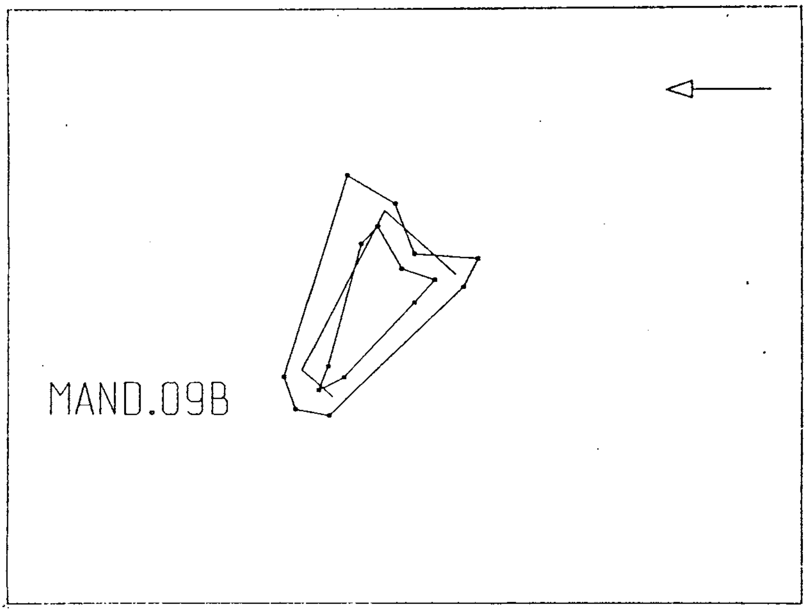
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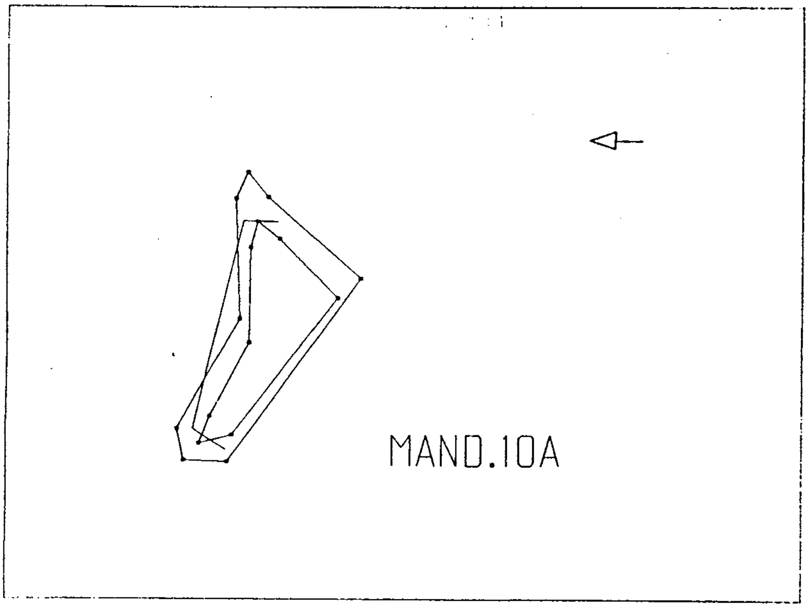
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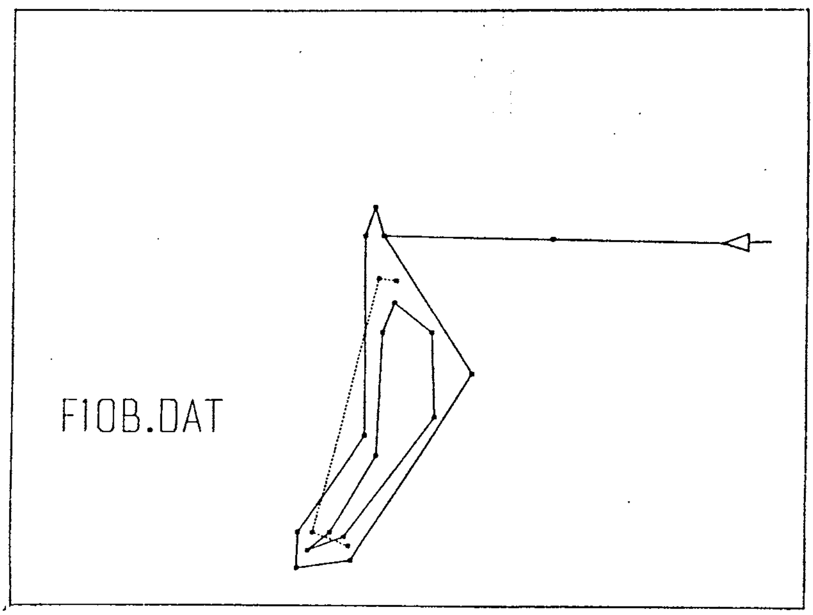
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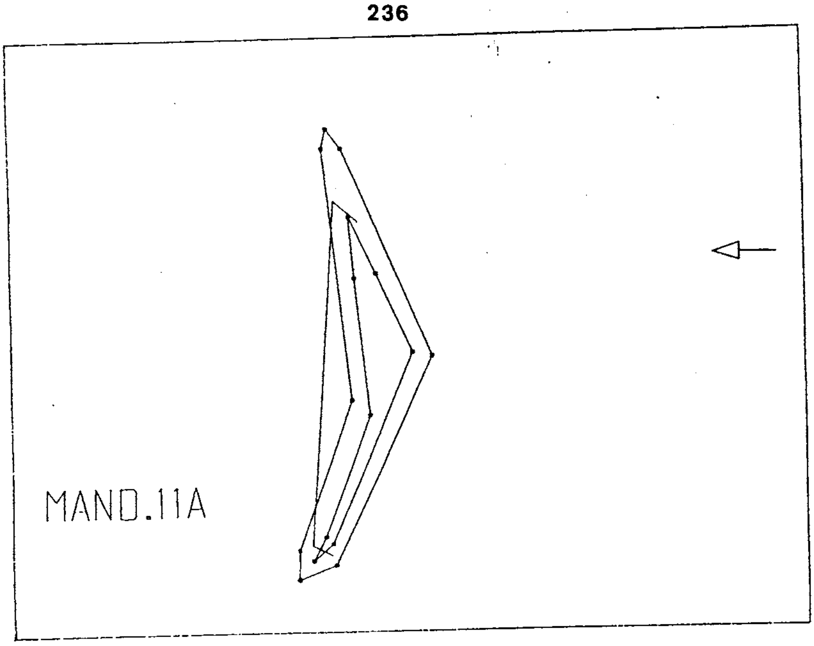
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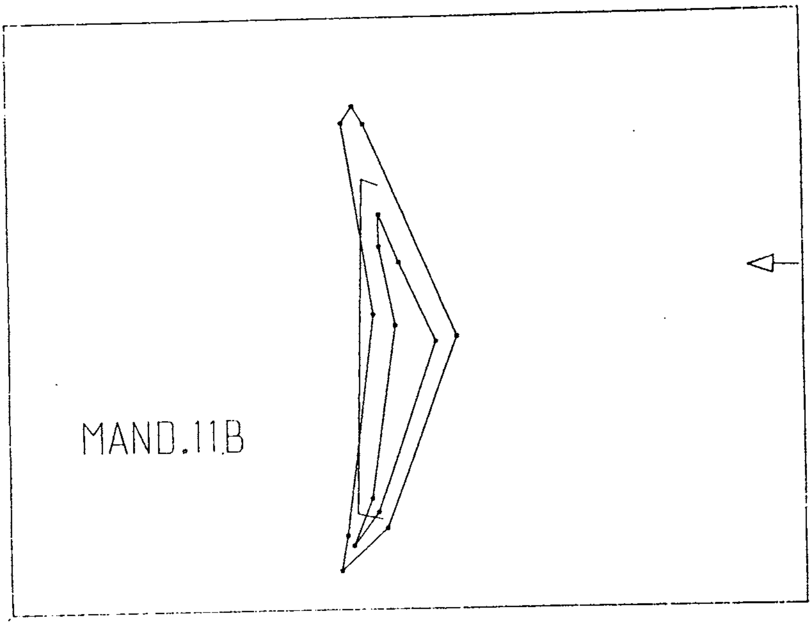
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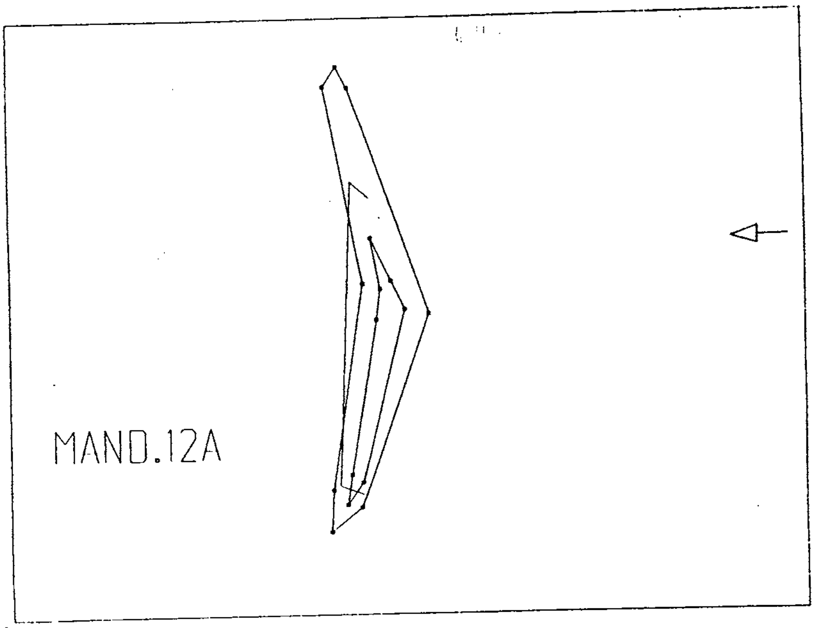
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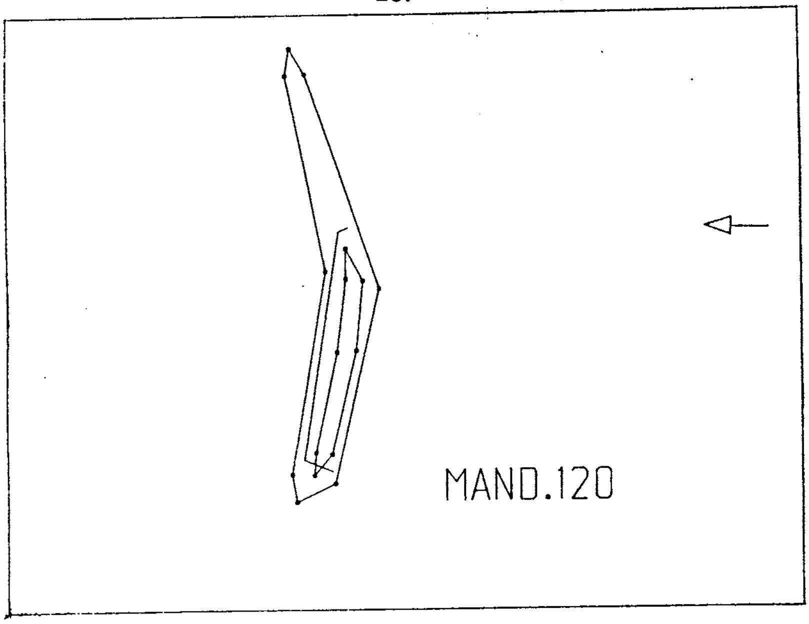
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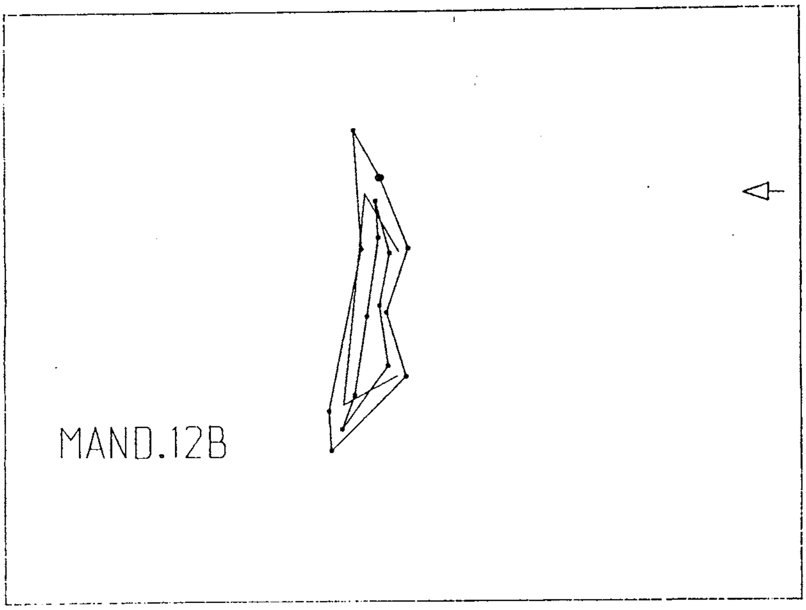
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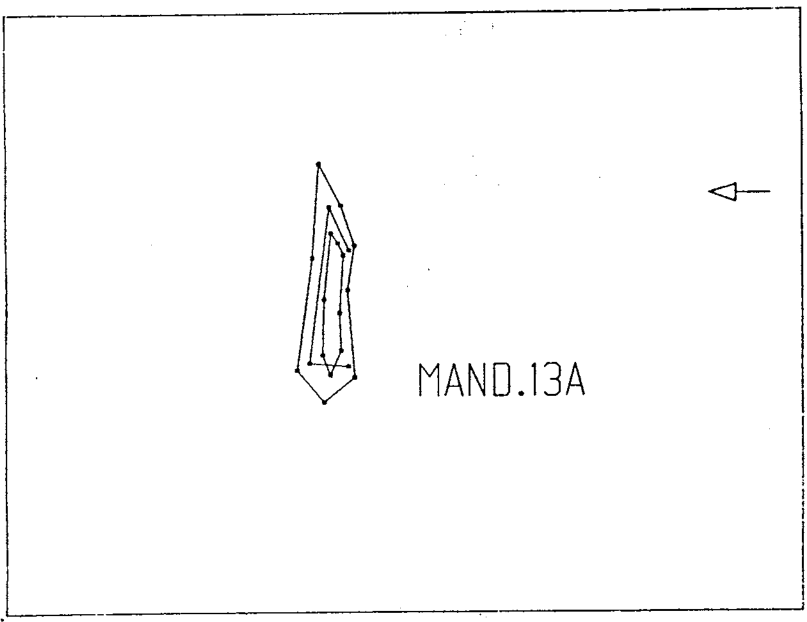
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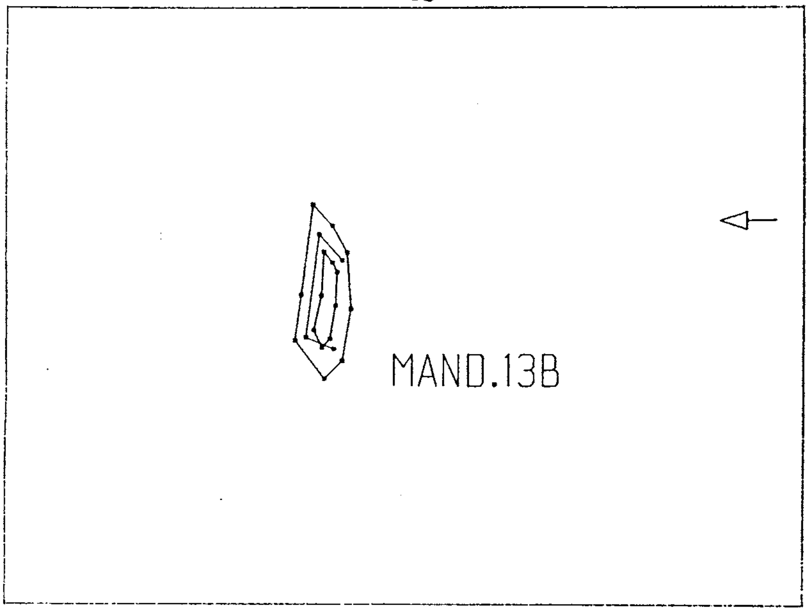
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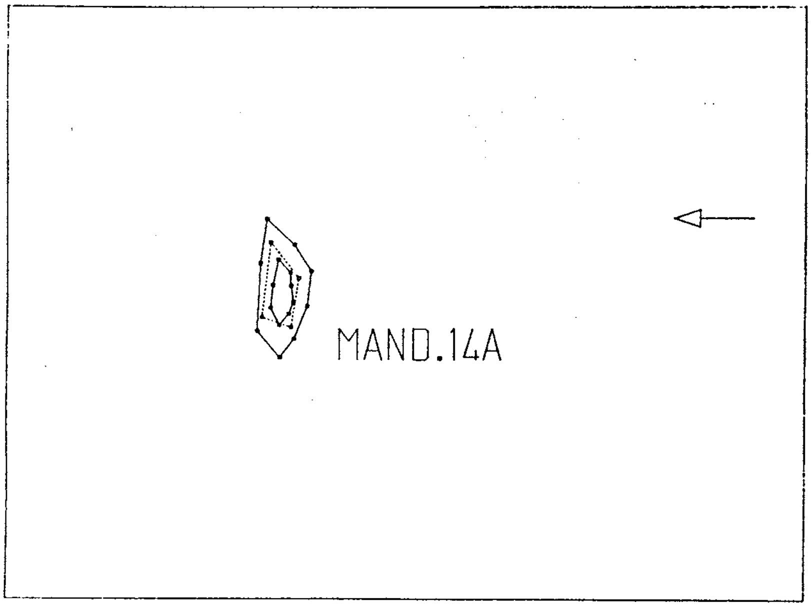
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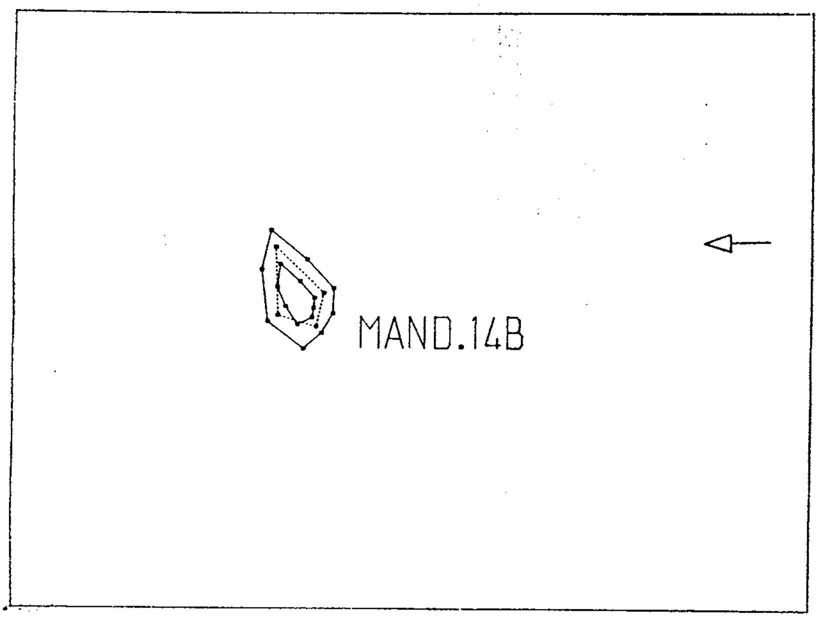
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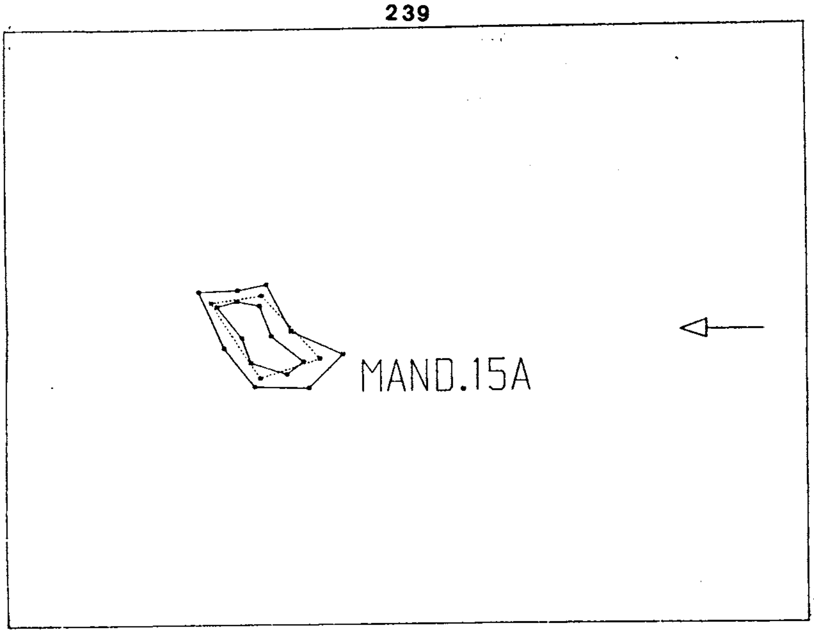
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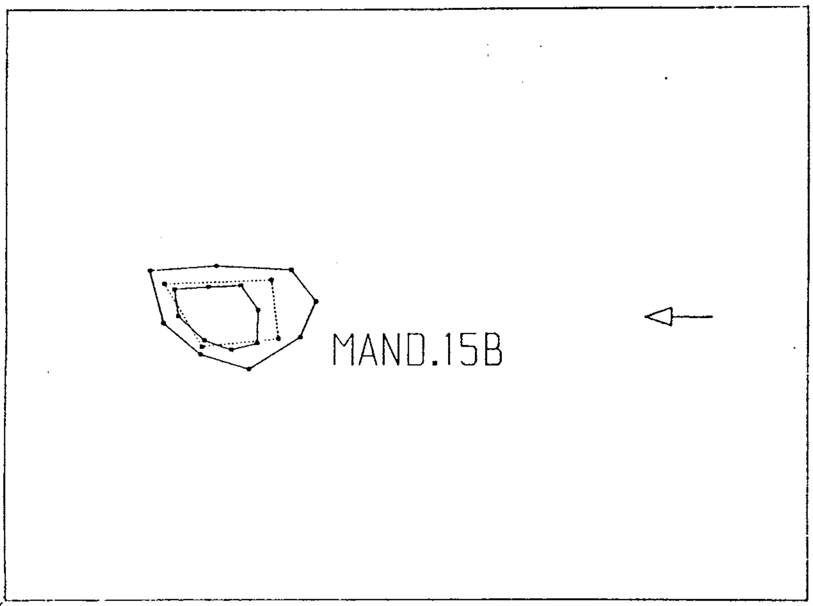
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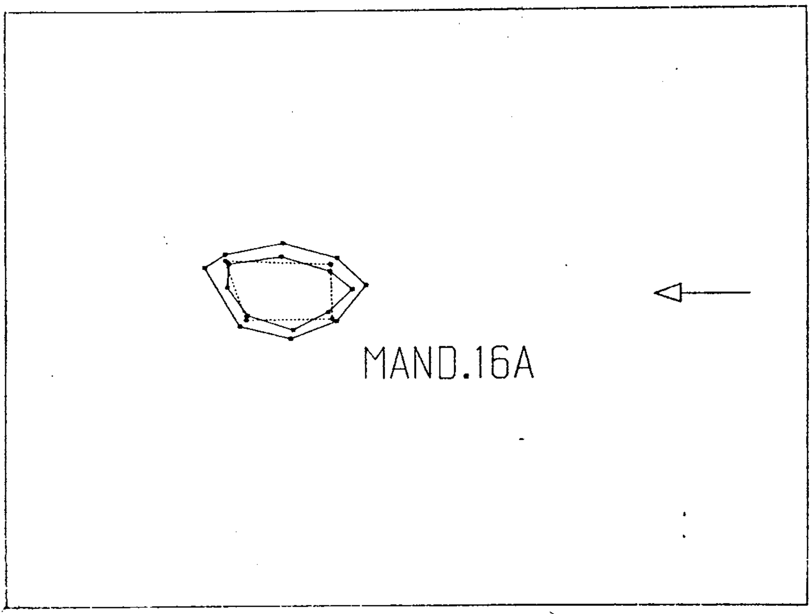
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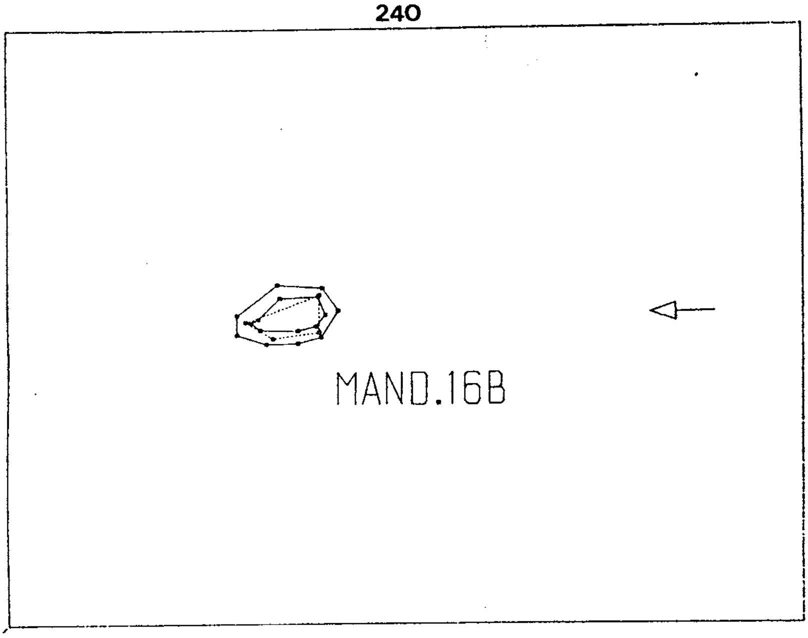

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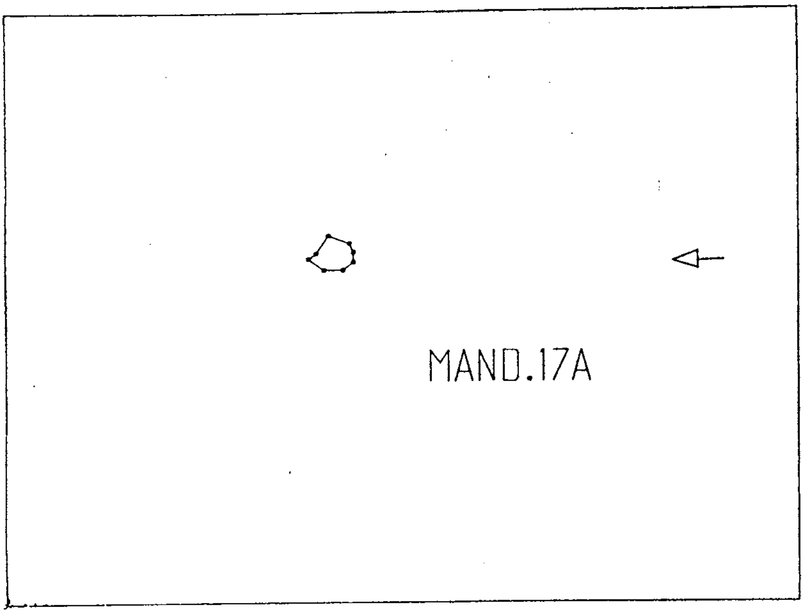


MAND.17A

APPENDIX II

General published anatomic data on the muscles of the mastication.

TABLE A		
MUSCLE	ORIGIN	INSERTION
BUCCINATOR	ALVEOLAR PROCESSES OF MAXILLAE AND MANDIBLE	ORBICULRIS ORIS FIBRES
* MASSETER	ZYGOMATIC PROCESS OF MAXILLA, ZYGO-MATIC ARCH	RAMUS, ANGLE AND CORONOIC PROCESS OF MANDIBLE
MENTALIS	MENTAL SYMPHYSIS	SKIN OF CHIN AND LOWER LIP
* PTERYGOID LATERALIS	GREAT WING OF SPHENOID, LATERAL PTERYGOID LAMNIA	CONDYLOID PROCESS
* PTERYGOID MEDIALIS	LATERAL PTERYGOID LAMNIA, PALATINE MAXILLA	RAMUS ANDANGLE OF MANDIBLE
RISORIUS	BUCCINATOR FASCIA	SKIN OF CORNERS OF MOUTH
* TEMPORALIS	TEMPORAL FOSSA	CORONOID AND RAMUS, MANDIBLE

TABLE B		
MUSCLE	GENERAL LOCATIION	ACTION
BUCCINATOR	CHEEK OF FACE	COMPRESSES FACE
* MASSETER	SIDE OF MANDIBULAR	ELEVATES MANDIBLE
MENTALIS	CHIN	DEPRESSES LIP, WRINKLES CHIN
* PTERYGOID LATERALIS	MEDIAL TO RAMUS OF MANDIBLE	PROTRUDES AND OPENS JAW
* PTERYGOID MEDIALIS	MEDIAL TO LATERALIS	CLOSES JAW
RISORIUS	LATERAL TO MOUTH	PULLS MOUTH LATERALLY
* TEMPORALIS	LATERAL SKULL	CLOSES JAWS

TABLE 1

Average length (mm) of muscles of mastication (SCHUMACHER, 1961).

	MALE	FEMALE
MED.PTERY.	17.7	14.9
MASSETER	26.7	25.2
TEMPORALIS	36.1	31.5
LAT.PTERY.	22.8	21.5

TABLE 2

Average weight (gr) of muscles of mastication (SCHUMACHER, 1961).

	WET WGHT.	%	DRY WGHT.	%
MED.PTERY.	3.13	11.1	0.92	11.1
MASSETER	7.85	28.0	2.35	28.1
TEMPORALIS	12.89	45.8	3.82	45.7
LAT.PTERY.	4.24	15.1	1.26	15.1

TABLE 3

Mean values(x), standard deviations(sd), and range of variation for the mandibular dimensions reported by Ringqvist (1973).

VARIABLE	X	S.D.	RANGE
Max. incisor bite force, kg.	29.9	6.0	20.5-45.7
Max. molar bite force, kg.	47.7	9.4	30.8-69.3
Overjet, mm	2.5	1.6	1.0-5.5
Overbite, mm	3.3	1.5	1.0-7.5
Max. opening, mm	54.8	4.7	41.0-62.0
Lateral movement, left, mm	10.6	1.7	6.5-14.0
Lateral movement, right mm	10.6	1.9	7.0-16.0
Protrusion, mm	8.2	2.1	4.0-12.0
Length of the mandibular base, mm	114.7	4.8	106.5-125.6
Length of the mandibular body, mm	80.5	5.5	68.0-92.0
Height of the ramus, mm	51.2	4.9	41.6-59.2
Height of the mand.alveolar proc., mm	29.7	2.5	27.2-34.1

TABLE 4

Reported average maximum opening (mm)

	MALE	FEMALE
Posselt (1952)	43.4	
Navakari (1960)	55.7	
Travell (1960)	59.0	53.0
Posselt (1962)	60.0	50.0
Sheppard (1965)	46.9	
Ringqvist (1973)	54.8	
Bosman (1974)	54.4	53.6
Agerberg (1974)	58.6	53.3
Rosenbaum (1975)	44.9	
Rieder (1978)	40.0-60.0	35.0-55.0

TABLE 5

Reported maximum bite forces (both sides) (N)

	PRUIM	RANGE	OSBORN
First PREMOLAR P_1	633	(386-908)	833
First MOLAR M_1	965	(609-1308)	1029
Second MOLAR M_2	756	(403-1206)	----
Third MOLAR M_3	---		1715

APPENDIX III

General published biomechanics data on the muscles of the mastication.

TABLE 6

Physiological cross sectional area (cm^2) of the facial muscles:

	SCHUMACHER	PRUIM	GRANT	BRYCE WEBER FICK CARLSOO	GYSI	BUCHNER	THIS WORK
MED.PTERY.	1.97		1.43	4.0	2.3	1.5	2.0
		5.3					
MASSETER	3.02		2.53	7.5	3.87	2.75	3.0
Anterior		2.6	1.65	5.19	–		
TEMPORAL.	3.81					3.45	3.8
Posterior		1.6	1.65	3.05	4.3		
OPENERS	–	1.0*	–	–	–	–	1.4
LAT.PTERY.	1.83	2.1	–	–	–	1.5	1.8

*Only anterior digastric was measured ($0.8cm^2$)

TABLE 7

Averaged calculated maximum muscle forces (N) (single side)

	HONEE (1970)	CARLSOO (1952)	SCHUMACHER (1961)*	PRIUM (1980)	OSBORN (1985)	THIS WORK
MED.PTERY.	–	299±46	190		254	191
	–			639±176		
MASSETER	–	614±107	340		450	340
Anterior	–	519±102		362±65	264	
TEMPORAL.			420			528
Posterior	–	305±102		197±26	323	
OPENERS	–	–		115±40	107	155
LAT.PTERY.	100±16.5	525	175	378±106	382	378

* Reported by Pruim et al.(1980)

TABLE 8

Average Maximum Muscle Tension (Maximum force exerted by a muscle to its physiological cross section)(N/m^2)

JOHNSON (1903)	$1.0x10^6$
FICK (1910)	$1.0X10^6$
MORRIS (1948)	$0.9X10^6$
CARLSOO (1952,1956)	$1.1x10^6$
HETTINGER (1961)	$0.4X10^6$
IKAI & FUKUNAGA (1968)	$0.7X10^6$
PRUIM (1980)	$1.4x10^6$
THIS WORK	$1.47x10^6$

TABLE 9
Muscle moment arms in (cm):

	BARBANEL (1972)	(1973) GRANT			**	PRUIM (1980)
			I C R*		C. Cent.	
		CLOSE	CLOSE	OPEN	OPEN	
MED.PTERY.	2.3	2.4	4.5	0.45	2.2	3.5
MASSETER	2.7	3.3	5.4	0.95	2.95	4.1
Anterior TEMPORALIS	2.6	3.1	5.8	5.2	3.4	3.3
Posterior TEMPORALIS		1.9	3.5	6.3	1.5	2.1
OPENERS	-	-	-	-	-	10.4
LAT.PTERY.	0.0	0.0	0.0	0.0	0.0	0.0

* ICR: Instantaneous centre of rotation.
** Condyle center

TABLE 10

Angular Directions (Degrees)

	PRUIM*	OSBORN	THROCKMORTON	THIS WORK**
MED.PTERY.	70.3	55.,67.	-	80.0
MASSETER	64.1	72.,75.,87.,91.	66.	80.0
Anterior	99.5			
TEMPORALIS		98.,129.,144.	112.	113.4
Posterior	139.8			
OPENERS	210.8	190.	-	210.8
LAT.PTERY.	0.0	10.,332.,356.	-	348.0

* Measured when condyle was located opposite the tubercle and not in the articular fossa.
** During clench.

APPENDIX IV

General published data on the mechanical properties of the mandibular bone.

TABLE 11

Various published Young's modulus (MPa) values for enamel, dentin, and PDL.

	ENAMEL	DENTIN	PDL
GUPTA (1972)	--	2.06×10^4	--
THRESHER (1973)	4.1×10^4	1.37×10^4	0.13×10^4
NORTON (1974)	--	2.06×10^4	--
WIDERA (1976)	--	2.07×10^4	0.69×10^2
KITOH (1977)	5.88×10^4	1.07×10^4	--
KNOELL (1977)	--	2.08×10^4	--
WEINSTEIN (1980)	--	2.07×10^4	0.69×10^2
GOEL (1976)	4.0×10^4	1.33×10^4	--

TABLE 12

Various published Young's modulus (MPa) values.

	CORTICAL BONE	ALVEOLAR BONE	CANCELLOUS BONE
GRENOBLE (1972)	1.79×10^4	--	--
NORTON (1974)	1.03×10^4	1.03×10^4	--
WIDERA (1976)	1.03×10^4		1.29×10^3
KITOH (1977)	0.98×10^4		0.5×10^3
KNOELL (1977)	1.79×10^4	1.04×10^4	
WEINSTEIN (1980)	1.37×10^4	--	--
LAVERNIA (1981)	1.37×10^4		0.69×10^3
BLACK (1973)	Wet $6.9\text{-}10.4 \times 10^3$ Dry $10.7\text{-}18.8 \times 10^4$		
REILLY (1974)	1.7×10^4		
EVANS (1973)	1.9×10^4		
SVENSSON (1977)	1.72×10^4		0.32×10^3
VICHNIN (1986)	1.15×10^4		0.325×10^3

TABLE 13

Variations in Young's moduli (MPa) values according to intermedullary cancellous bone percentages.

	KNOELL (1977)	WEINSTEIN (1980)
1%	0.2×10^3	0.005×10^3
10%	--	0.5×10^3
15%	2.7×10^3	--
25%	--	1.5×10^3
30%	5.4×10^3	--
40%	7.2×10^3	--
50%	--	4.0×10^3
65%	11.7×10^3	--
75%	--	7.0×10^3
90%	--	11.0×10^3

TABLE 14

The elastic moduli of bone (GPa)

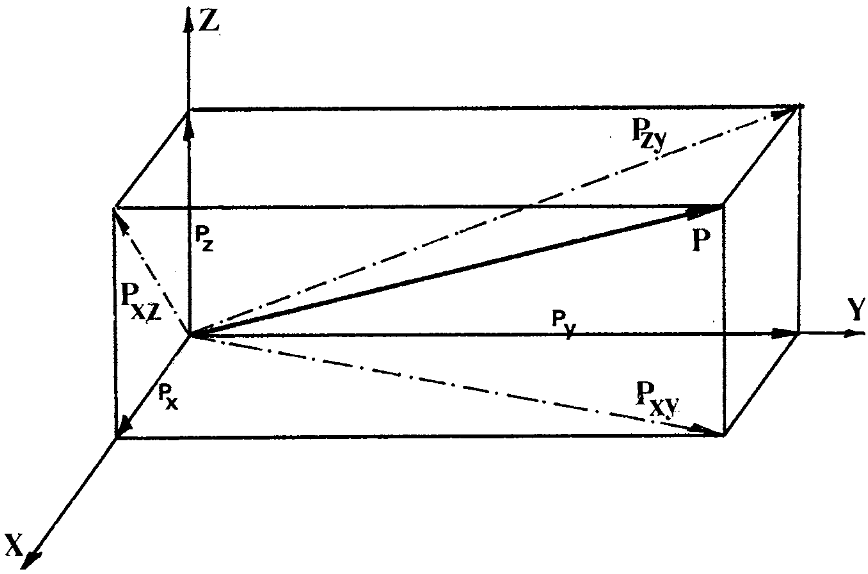
	Swanson		Van Burskirk	Vichnin et al.
	WET	DRY	(1981)	(1986)
LONGITUDINAL	8.78	12.1	21.5	17.0
TRANSVERSE	4.21	6.29	14.4	11.5
RADIAL	3.8	6.4	13.0	-

APPENDIX V

Details of the force analysis of the muscles of mastication. Various previously published data and anatomic observations were utilised to determine the magnitude and the direction of the muscles.

ESTIMATED MUSCLE FORCE MAGNITUDE (N) AND DIRECTIONS
(ONE SIDE ONLY)
A. CLENCH

	MASSETER	TEMPORALIS A	TEMPORALIS B	MEDIAL PT	LATERAL PTERYGOID	OPENERS*
P	340.0	264.2	264.2	191.4	378.1	155.0
P_x	141.9	50.6	50.6	-82.9	-215.4	0.0
P_y	-53.6	+102.9	+102.9	29.7	-307.6	+133.2
P_z	304.3	238.1	238.1	170.0	-43.2	-79.3
P_{xy}	151.7	114.6	114.6	-88.1	375.5	0.0
P_{zy}	309.0	259.3	259.3	172.5	310.6	155.0
P_{xz}	335.8	243.4	243.4	9.1	219.7	0.0



ESTIMATED MUSCLE FORCE MAGNITUDE (N) AND DIRECTIONS
(ONE SIDE ONLY)
B. WIDE OPENING MOVEMENT

	MASSETER	MEDIAL PTERYGOID	LATERAL PTERYGOID	OPENERS
P	129.0	65.36	377.6	155.0
P_x	44.7	-24.5	-197.2	0.0
P_y	0.0	0.0	-218.6	135.6
P_z	121.2	60.6	156.1	-75.1
P_{xy}	-	-	343.8	-
P_{xz}	-	-	322.0	155.0
P_{xz}	129.0	65.36	251.5	-

ESTIMATED MUSCLE FORCE MAGNITUDE (N) AND DIRECTIONS.
(ONE SIDE ONLY)
C. PROTRUSION

	MEDIAL PTERYGOID	LATERAL PTERYGOID	OPENERS*
P	196.4	377.6	155.0
P_x	-74.46	-197.2	0.0
P_y	0.0	-281.6	135.6
P_z	181.79	156.1	-75.1
P_{xy}	-	343.8	-
P_{zy}	-	322.0	155.0
P_{xz}	196.4	251.5	-

* Openers are: Digastrics, mylohyoid and geniohyoid muscles.

A. EQUILIBRIUM EQUATIONS:

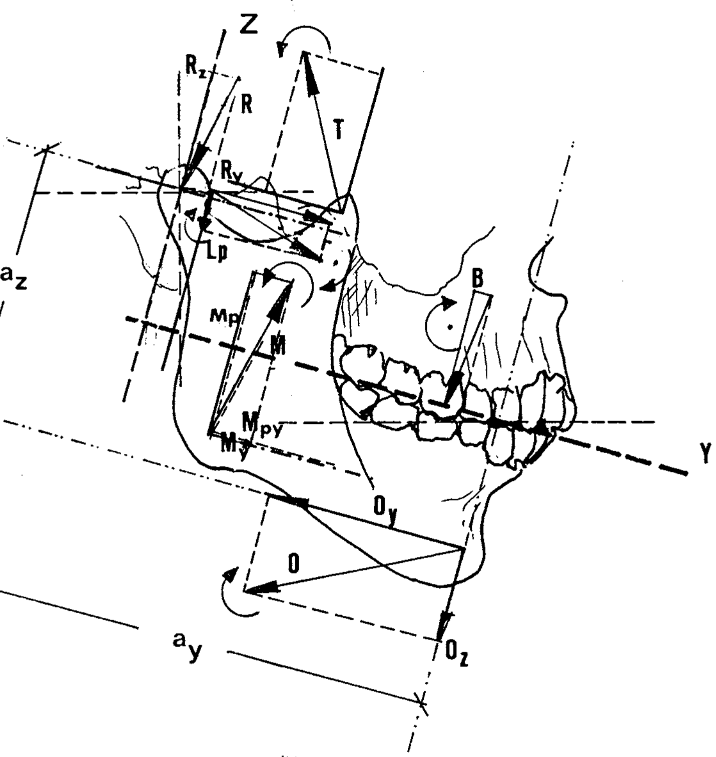
$$\overline{P} = \sqrt{(P_{ix})^2 + (P_{iy})^2 + (P_{iz})^2}$$

P_{ix} = sum of projections of all forces on x axis,

P_{iy} = sum of projections of all forces on y axis,

P_{iz} = sum of projections of all forces on z axis.

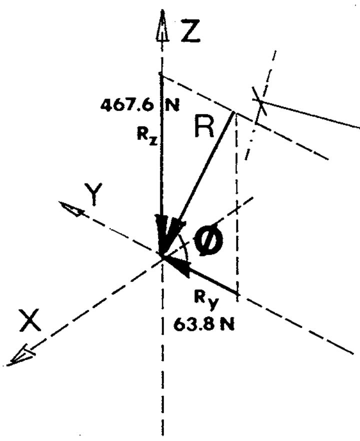
$$M_x = P_y a_z + P_z a_y$$



$$P_{iy} = 0$$

$$P_{iz} = 0$$

$$M_x = 0$$



REACTION FORCE AND MUSCLE FORCE CALCULATIONS

A. CLENCH

MUSCLES	P_x	P_y	P_z	a_y	a_z	M_x
MASSETER A	141.9	-53.6	304.3	-30.0	-30.25	10750.4
MASSETER B	-141.9	-53.6	304.3	-30.0	-30.25	10750.4
TEMPORALIS A	50.6	102.9	238.1	-30.0	-16.8	8871.7
TEMPORALIS A	50.6	102.9	238.1	-30.0	-16.8	8871.7
TEMPORALIS B	-50.6	102.9	238.1	-30.0	-16.95	8887.1
TEMPORALIS B	-50.6	102.9	238.1	-30.0	-16.95	8887.1
MEDIAL PTERY. A	-82.9	-29.7	170.0	-30.0	-29.4	5973.18
MEDIAL PTERY. B	82.9	-29.7	170.0	-30.0	-29.4	5973.18
LATERAL PTER.A	-215.4	-307.6	-43.2	-4.0	5.1	-1741.56
LATERAL PTER.B	215.4	-307.6	-43.2	-4.0	5.1	-1741.56
OPENERS	0.0	133.2	-79.3	-81.0	-24.7	-9713.34
TOTAL	0.0	-237.0	1735.3			55768.3
JOINT REACTION A	0.0	63.8	-467.6	0.0	0.35	
JOINT REACTION B	0.0	63.8	-467.6	0.0	0.35	
BITE FORCE A	0.0	54.7	-400.0	70.0	14.6	
BITE FORCE B	0.0	54.7	-400.0	70.0	14.6	

REACTION FORCE AND MUSCLE FORCE CALCULATIONS

B. WIDE OPEN

MUSCLE	P_x	P_y	P_z	a_y	a_z	M_x
MASSETER A	44.7	0.0	121.2	-30.0	-30.25	3636.0
MASSETER B	-44.7	0.0	121.2	-30.0	-30.25	3636.0
MEDIAL PTER A	-24.5	0.0	60.6	-30.0	-29.4	1818.0
MEDIAL PTER B	-24.5	0.0	60.6	-30.0	-29.4	1818.0
LAT. PTER A	-197.2	-281.6	156.1	-4.0	5.1	-811.7
LAT. PTER B	197.2	-281.6	156.1	-4.0	5.1	-811.7
OPENERS	0.0	135.6	-75.1	-81.0	-24.7	-9432.4
TOTAL	0.0	-427.6	600.7			-147.8
JOINT REACTION	0.0	213.0	-300.3	0.0	0.35	
JOINT REACTION	0.0	213.0	-300.3	0.0	0.35	