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THE UNIVERSITY OF ALBERTA

A Mechanical Analysis of the Temporomandibular Joint

by

Andrew Stuart Hay

A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES AND RESEARCH

IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE

OF MASTER OF SCIENCE

MECHANICAL ENGINEERING

EDMONTON, ALBERTA

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Supervisor

Date: April 19, 1985

ABSTRACT

The purpose of this investigation was to develop an *in vitro* model and a mathematical model of the human mandible to determine the effect of occlusal position, occlusal angle, and muscle forces upon occlusal loads and joint reactions. The two models agreed with each other and collaberated with most *in vivo* experimental findings and showed that (1) for unilateral occlusion in the molar region, contralateral TMJ loads exceeded ipsilateral TMJ loads; (2) as the point of occlusal contact was moved from the first molar distally, occlusal loads increased and TMJ loads decreased; (3) contralateral TMJ loads decreased as occlusal angle was changed from that of an unworn dentition to a worn dentition while ipsilateral TMJ loads increased and occlusal loads remained fairly constant.

ACKNOWLEDGMENT

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LIST OF ABBREVIATIONS

Ant. Temp.	= anterior temporalis
BFT	= bite force transducer
CFT	= condylar force transducer
Deep Massr.	= deep masseter
EMG	= electromyographic
Int. Ptery.	= internal pterygoid
Lat. Ptery.	= lateral pterygoid
M1	= first molar
M2	= second molar
M3	= third molar
Post. Temp.	= posterior temporalis
Sup. Massr.	= superficial masseter
TMJ	= temporomandibular joint

LIST OF NOMENCLATURE

Angles Class I Occlusion: a 'normal' occlusion.

Balancing Side: the opposite side of the mandible on which occlusion is occuring.

Bilateral: affecting both sides.

Bolus: a small round lump or mass.

Cathetometer Comparator: an optical device used to measure changes in angle.

Compact Bone: densely packed bone consisting largely of concentric lamellar osteons and interstitial lamellae.

Condyle: the rounded process at the end of a bone.

Contralateral: the opposite side.

Cranial: pertaining to the skull

Curve of Spee: anatomical curvature of the occlusal alignment of teeth beginning at the tip of the lower cuspid following the buccal cusps of the natural bicuspids and molars, continuing to the anterior border of the ramus.

Cusp: the elevations on the chewing surface of a tooth.

Dentition: relating to the arrangement of teeth in the mouth.

Electromyography: the electrical response of a muscle tissue to nerve stimulation.

Eminence: a raised area of bone.

Foramen Magnum: the opening at the base of the skull through which the spinal cord passes.

Force Vector: a vector which defines both the direction and the magnitude of the force.

Fossa: a cavity or small hollow.

Incisal: refering to the front teeth between the canines in either jaw.

Insertion: the point of attachment of a muscle to the part that it moves.

In Vitro: isolated from the living organism.

In Vivo: occuring within a living organism.

Ipsilateral: the same side

Ligament: a band of tough tissue connecting bones.

Mandible: the lower jaw.

Mastication: chewing or grinding of food.

Maxilla: the upper jaw.

Moment: the product of a force and its distance from a point of reference.

Occlusal Angle: the angle between the occlusal plane and the line of action of the occlusal force.

Occlusion: the way in which the upper and lower teeth fit together.

Origin: the less moveable of the two points of attachement of a muscle, usually to the more rigid part of the skeleton.

Orthoganal: at right angles.

Osseous: bony

Position Vector: a vector which defines the position of a point in relation to an origin.

Process: a projection or outgrowth of bone.

Reaction: an opposing force.

Statically Determinate: a problem which is solvable.

Strain Gauge: a wire foil which changes resistance in direct proportion to changes in strain.

Synovial Fluid: clear lubricating fluid secreted by the membranes of joint cavities.

Temperature Compensated: a circuit insensitive to changes in electrical resistance from changes in temperature.

Trabecular Bone: bone in which trabeculae are interconnected to form a latticework with interstices filled with connective tissue or bone marrow.

Unilateral: one side only.

Unit Vector: a vector whose length is unity.

Vertical Vernier: a measuring instrument which measures length vertically.

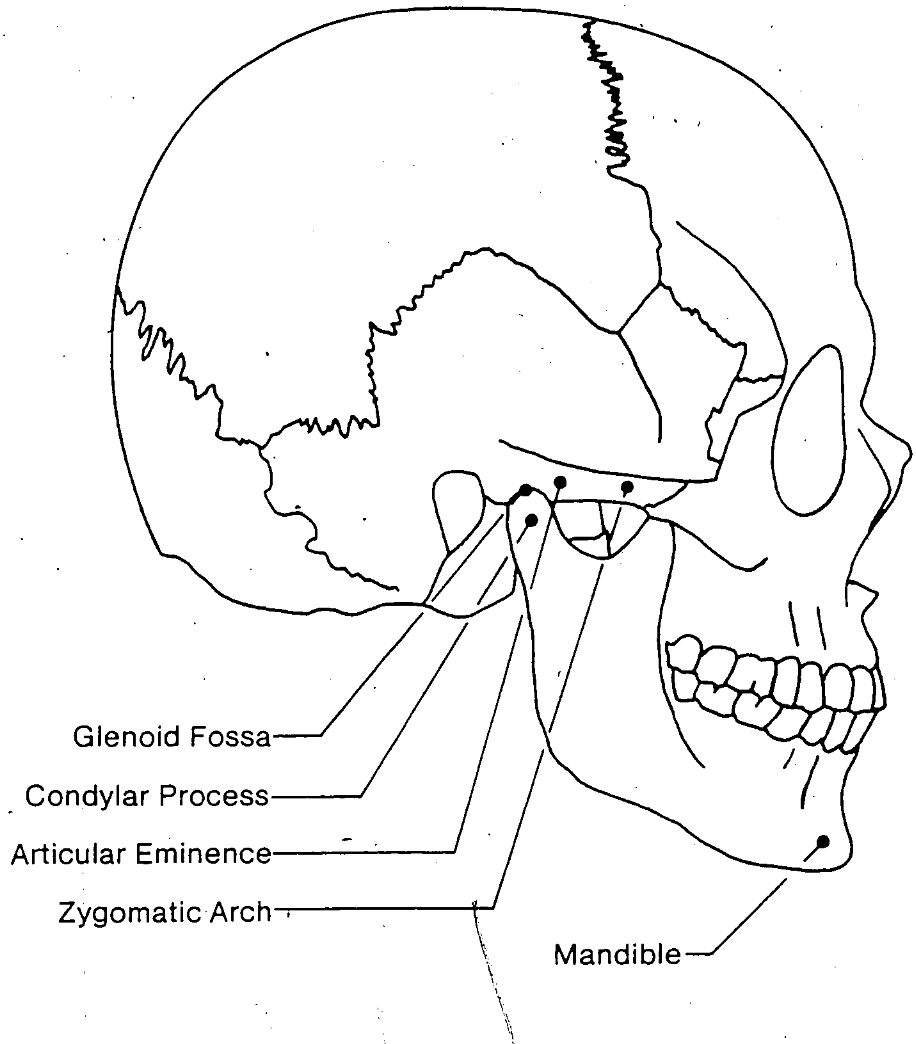
Wheatstone Bridge: a divided circuit used to determine changes in electrical resistance.

Working Side: the side of the mandible on which unilateral mastication is occuring.

1. INTRODUCTION AND ORAL ANATOMY

The human temporomandibular joint (TMJ) is comprised of soft tissue and osseous components. The soft tissue components of each joint include an articular disc, fibrous connective tissues, and four major ligaments. The osseous components include the condylar process of the mandible and the articular eminence and glenoid fossa of the temporal bone (figure 1). This joint, which is involved in the movements of the mandible, including speech, ingestion, and mastication, is perhaps the most complex joint in the body. Its' complexity arises from the ability of this joint to facilitate both rotation and translation, unlike any other human joint.

Recent studies have shown that the TMJ is often subject to dysfunction and disease [25, 26]. Mandibular dysfunction is a musculo-skeletal disorder that involves tissue injury [84]. These tissue injuries involve a spectrum of change from remodelling to degenerative joint disease within the joint. It has also been noted that hyperactivity of the muscles of mastication may damage these muscles leading to pain and altered muscle function [83]. In general, investigators of mandibular dysfunction have concentrated on the structure and function of the jaw including the structure of the TMJ, muscles of mastication, and the dental occlusion. The present study considers all of these areas but the focus will center around the TMJ components.



Lateral Aspect of Adult Human Skull

FIGURE 1

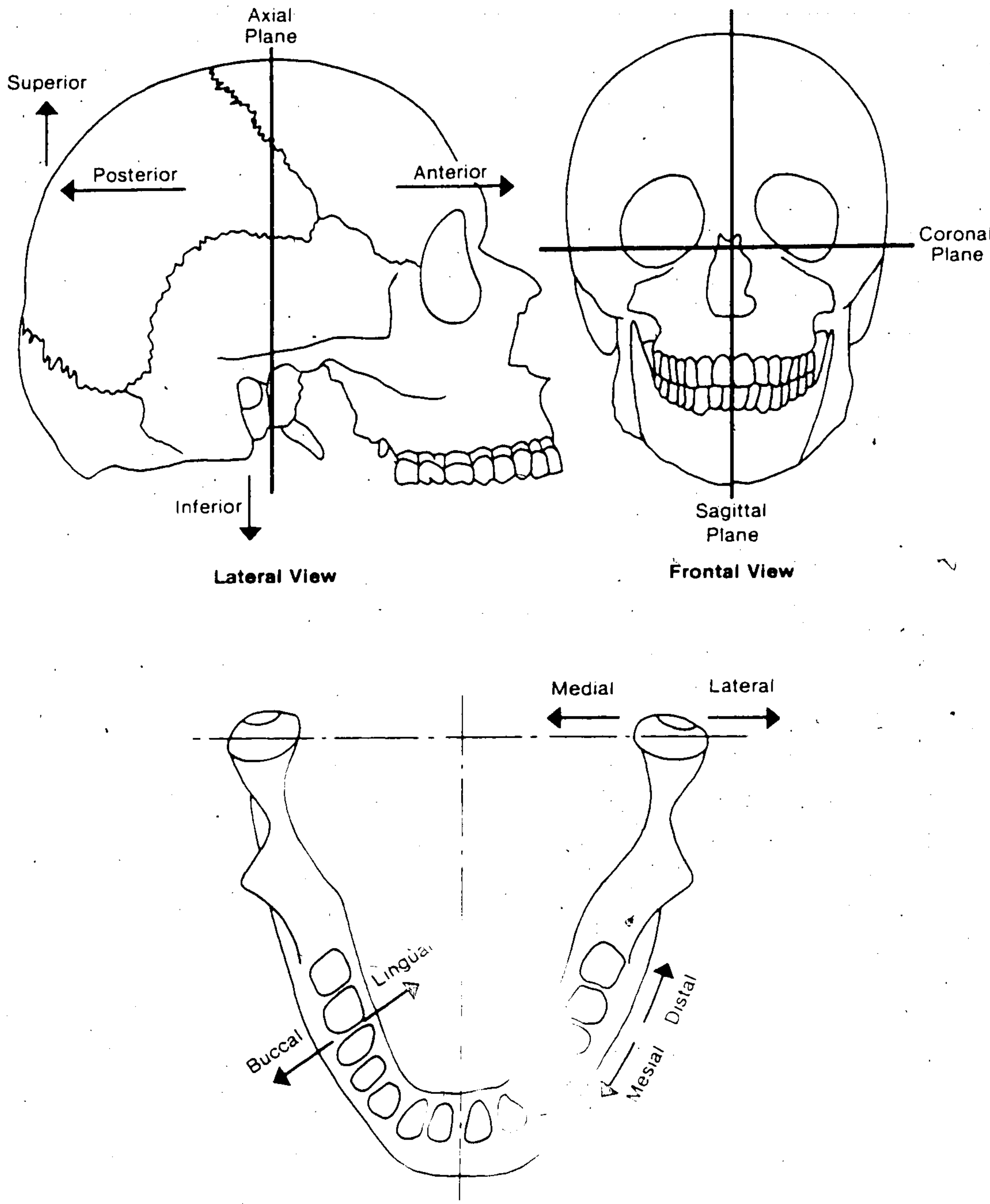
There has been increased evidence that a major factor involved in development of these clinical problems is joint force which exceeds the adaptive capabilities of the joint [61, 64, 65]. Clinical studies have also indicated a strong possibility that TMJ dysfunction is correlated to the loss of posterior teeth and dento-skeletal abnormalities such as mandibular asymmetries [20]. It is therefore hypothesized that the loss of posterior teeth or dento-skeletal abnormalities alter the resultant joint forces during normal mandibular function.

The purpose of this study was to develop both *in vitro* and mathematical models of the human masticatory system, which allowed examination of the effect of occlusal position (missing teeth), occlusal angle, and bite load upon joint reactions. The *in vitro* model was developed first, as it established the correct anatomical proportions. The mathematical model was then developed from measurements of the *in vitro* model. Once the two models were in agreement, the mathematical model was further utilized to investigate the effect of occlusal angle, muscle force, and occluding positions upon joint reactions.

1.1 ANATOMY

1.1.1 STRUCTURE

Refer to figure 2 for anatomical reference planes and directions. The human TMJ includes the condylar process, an

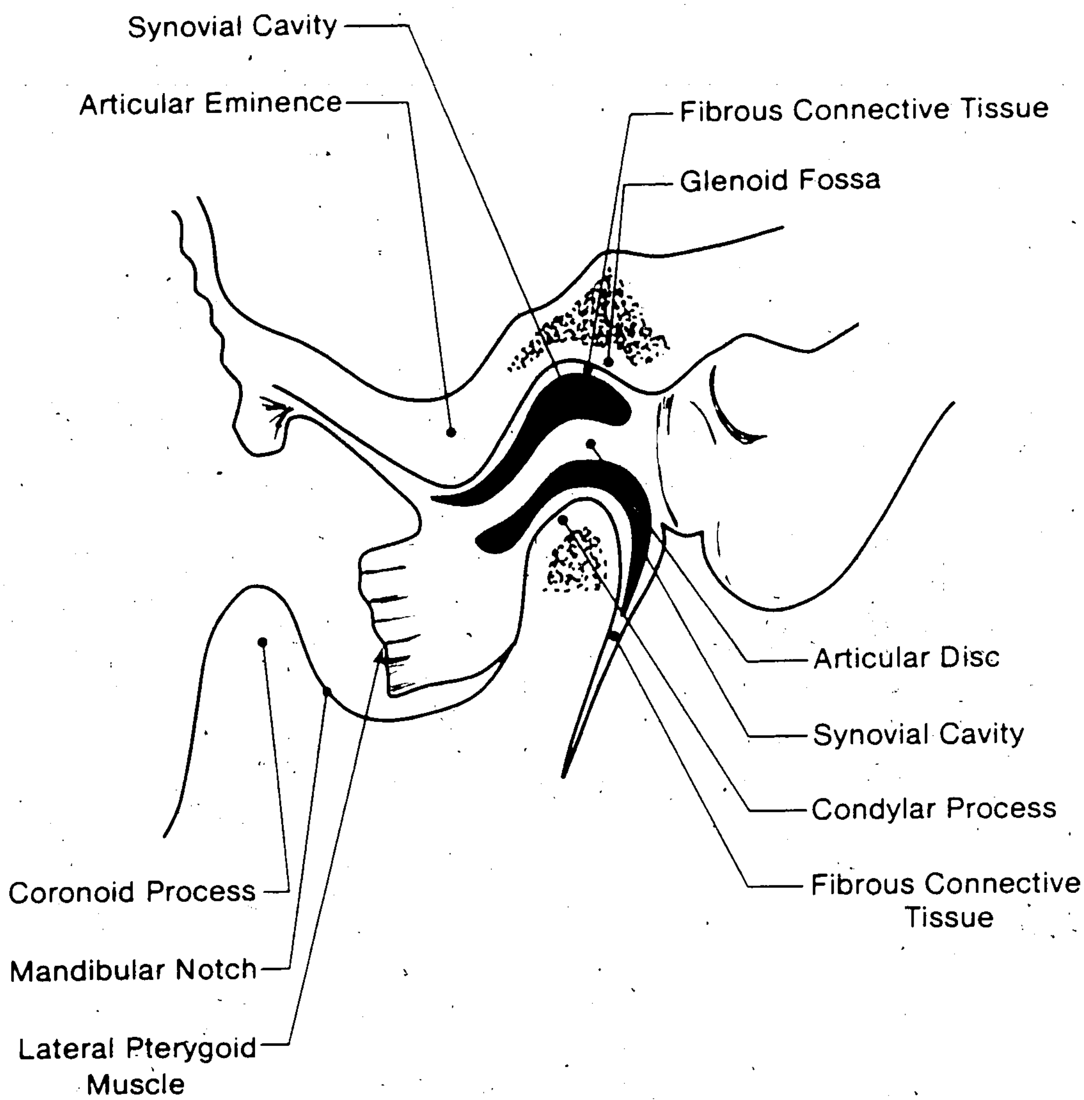


Anatomical Reference Planes and Directions

FIGURE 2

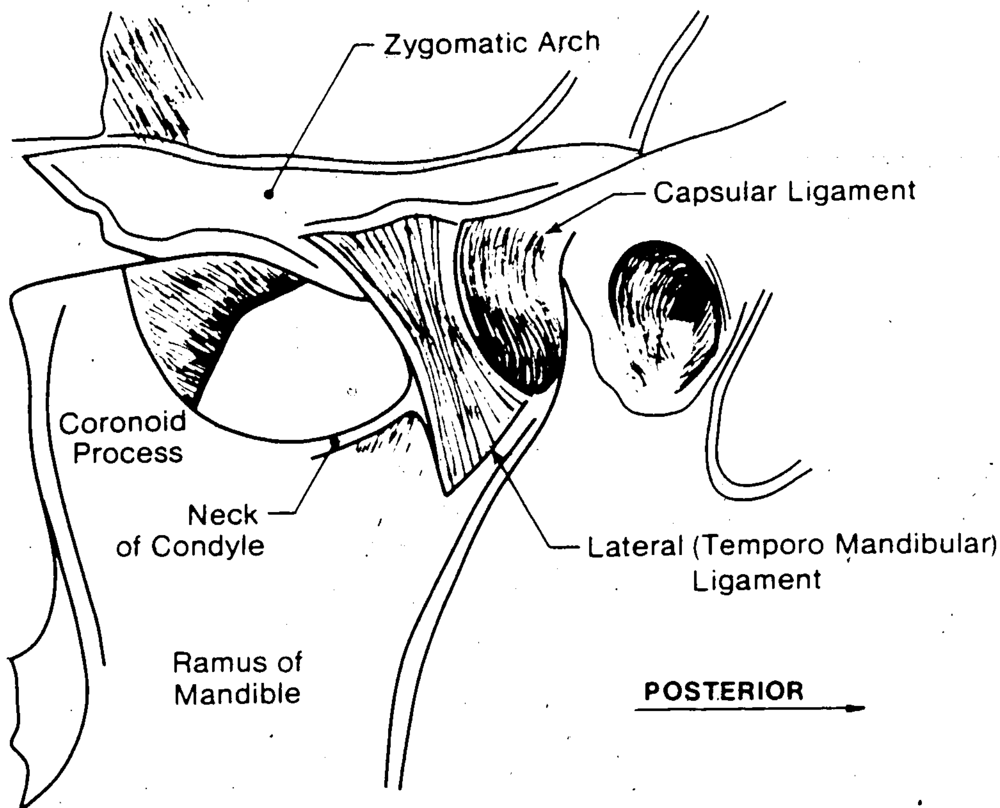
articular disc, the articular eminence and gleniod fossa of the temporal bone, and associated ligaments (figure 3). The fossa, eminence, and condyle are made of compact bone on their outer surfaces supported by an inner network of trabecular bone and are covered by a thin layer of fibrous connective tissue. The articular disc is a pad of fibrous connective tissue which is interposed between the condyle and the eminence/fossa structure, separating the joint into upper and lower compartments. Lubrication for the joint is provided by synovial fluid. The following ligaments encapsulate and stabilize the joint: the capsular ligament, the lateral ligament, the sphenomandibular ligament and the stylomandibular ligament (figure 4). Because of the unique structure of this joint, the condyle can be translated forward onto the eminence, rotated in the sagittal plane, and rotated in the coronal plane (figure 5).

The mandible (figure 6) is a structure made of an outer layer of compact bone, supported by an inner network of trabecular bone. One of its functions is to support teeth and the occlusal loads resulting from them. The teeth are held to the mandible in bony sockets by periodontal ligaments. During forceful occlusal contact, the forces are transferred from the teeth by the periodontal ligaments into the supporting structure of trabecular bone and from thence into the dense compact bone of the mandible [76, 77].

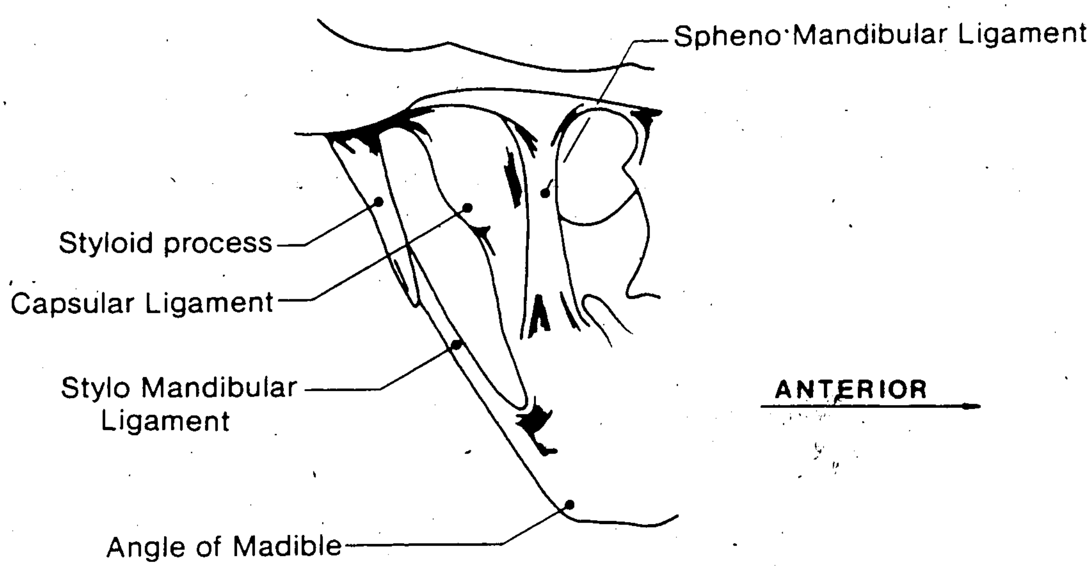


Lateral View of Temporomandibular Joint

FIGURE 3



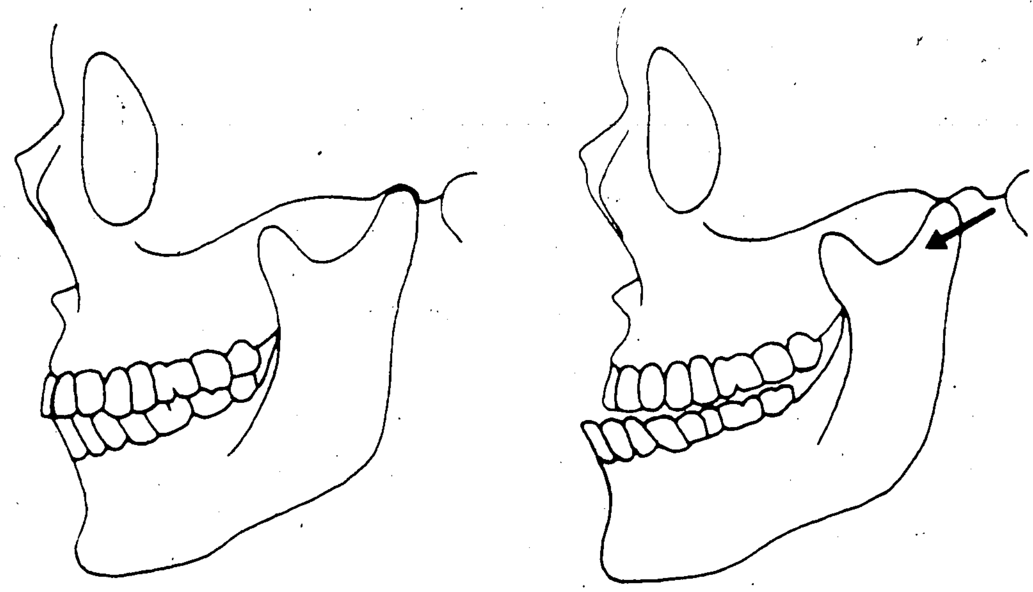
Lateral View of Temporomandibular Joint



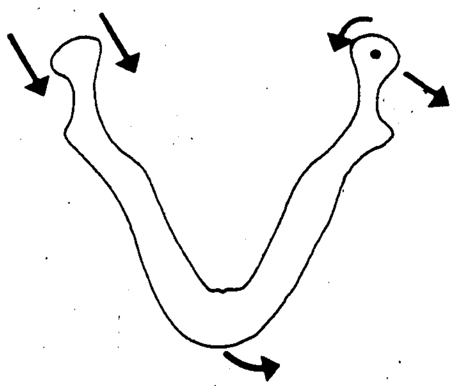
Medial View of Temporomandibular Joint

Ligaments

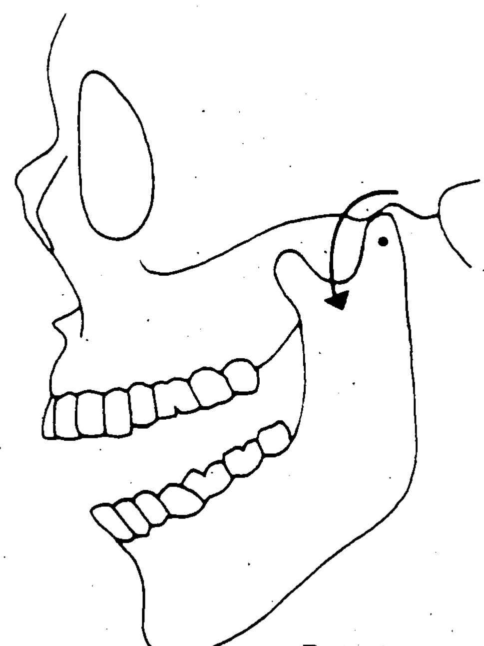
FIGURE 4



Forward Translation



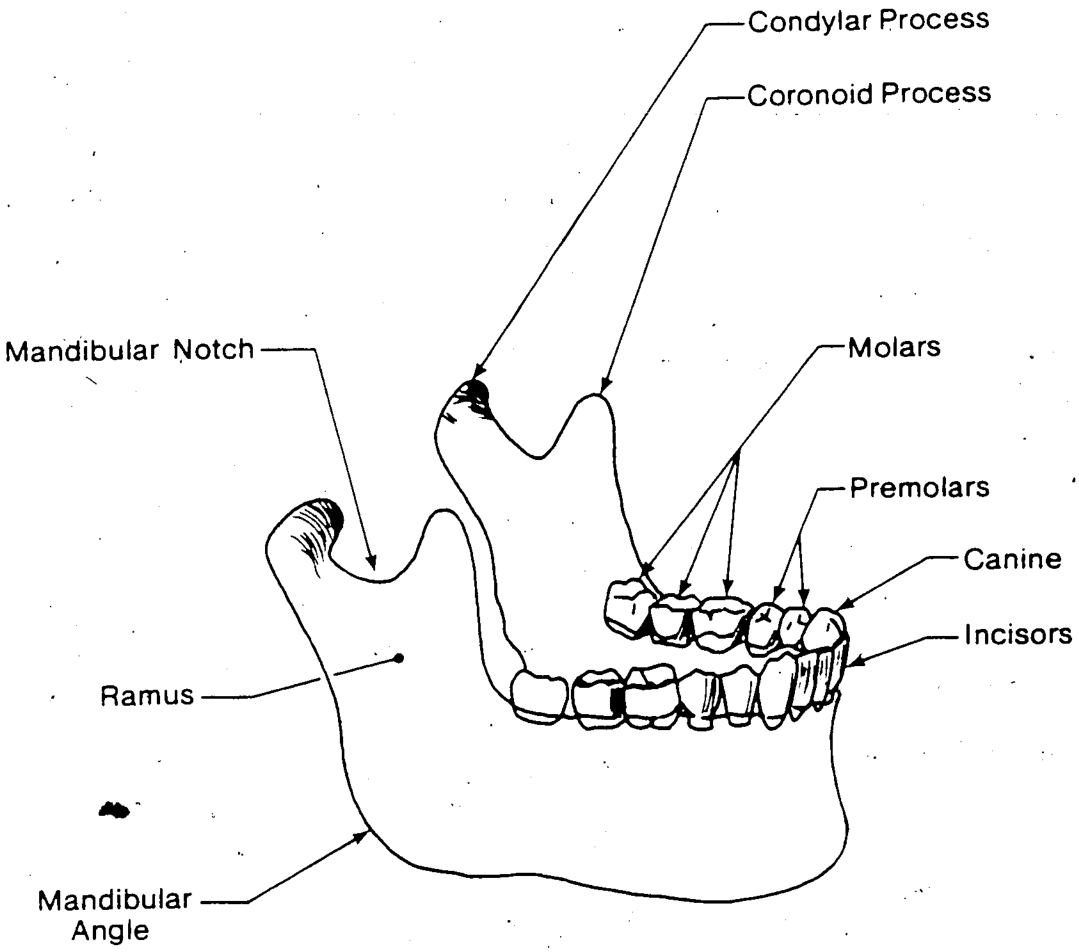
Rotation
(Left Condyle - Coronal Plane)



Rotation
(Sagittal Plane)

Condylar Movements

FIGURE 5



Mandible

FIGURE 6

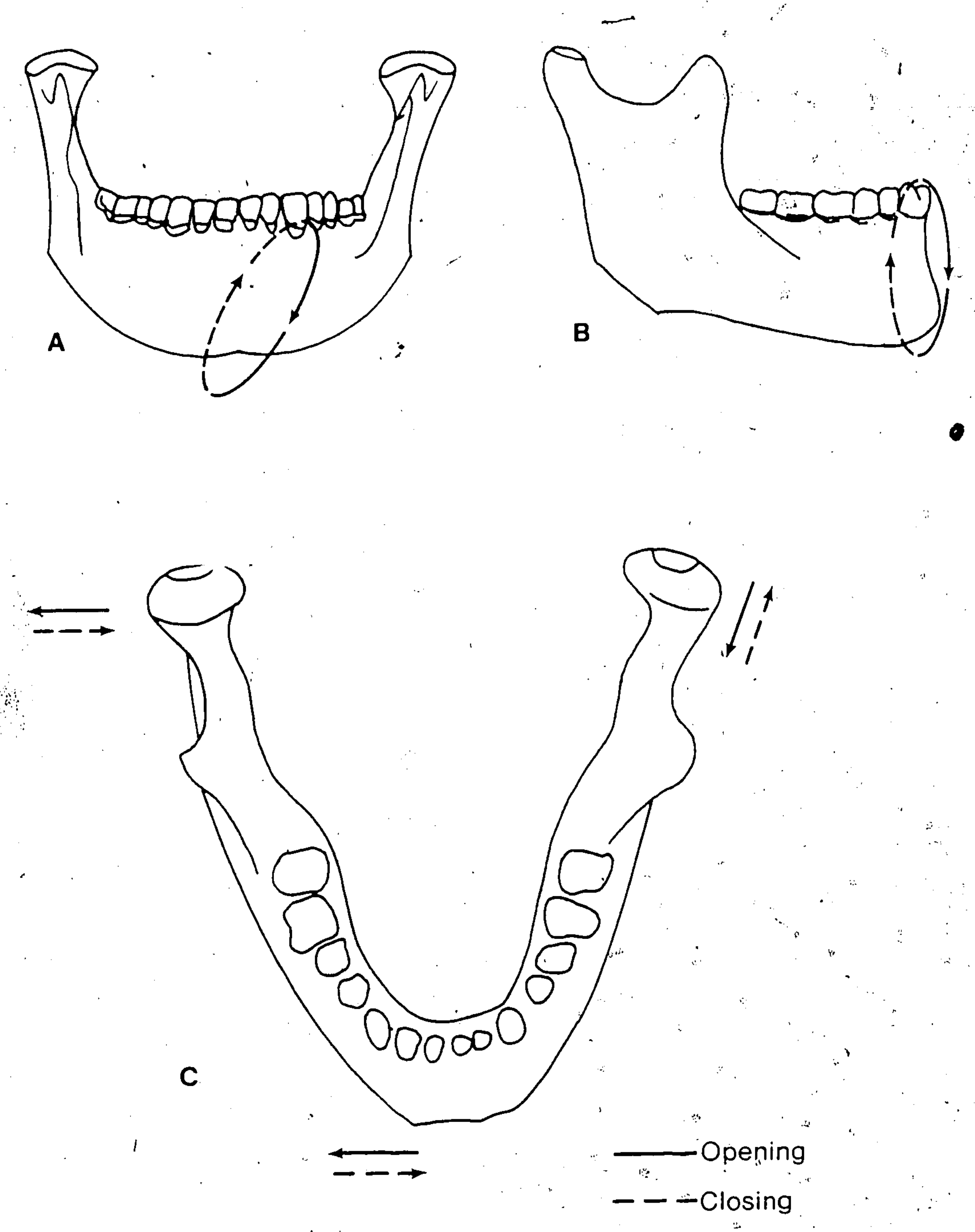
1.1.2 MOVEMENTS

The movements of the mandible can in general be categorized into two divisions: free movements and masticatory movements. Free movements include opening, closing, protrusion, retrusion, and lateral movements. Masticatory movements are those which involve forceful occlusal contact, such as incisal biting and grinding movements (mastication). In this study, only the forces developed in mastication are examined.

Mastication can be subdivided into three basic movements: opening, closing, and power strokes (figure 7). In the opening stroke, the molar teeth on the occluding side of the mandible are moved laterally and inferiorly. To accomplish this, the ipsilateral[1] condyle rotates and shifts laterally while the contralateral[2] condyle is translated anteroinferiorly and medially onto the eminence. The closing stroke follows, in which the occluding molar teeth are moved superiorly and medially into the bolus of food. To facilitate this movement, the ipsilateral condyle rotates in its fossa while the contralateral condyle is translated posterosuperiorly and laterally back into its fossa. The power stroke then occurs, in which occlusal contact takes place with much force, crushing and grinding the food

---

[1] ipsilateral refers to the same side of the mandible on which occlusion is occuring -for example: if occlusion is occuring in the left molar region, the ipsilateral condyle would be the left condyle. Another common word used for ipsilateral is "working".
[2] contralateral refers to the opposite side of the mandible on which occlusion is occuring. Another common expression for contralateral is "balancing".



Opening and Closing Strokes of Mastication
Unilateral Mastication – Right Side

FIGURE 7

object.

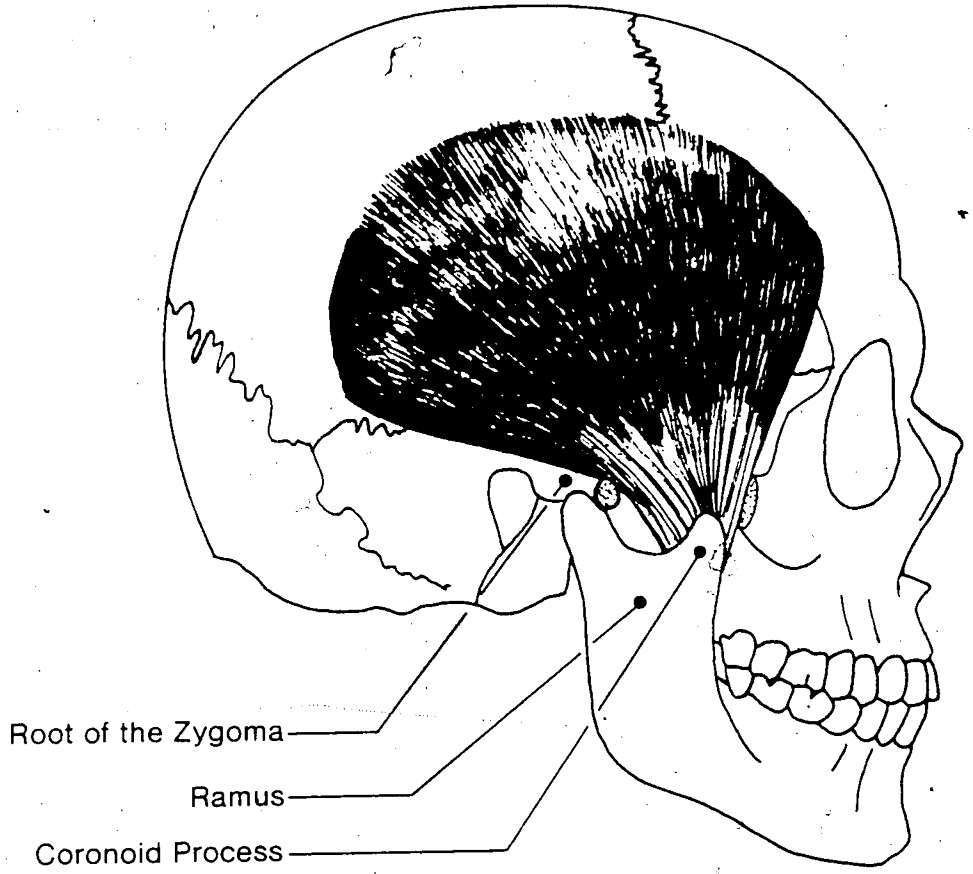
1.1.3 MUSCLES

The muscles involved in the power stroke of mastication are the deep and superficial masseter, the temporalis, the medial pterygoid, and the lateral pterygoid muscles. The temporalis muscle is fan-shaped and originates from the lateral surface of the temporal bone (figure 8). The muscle fibres converge and pass between the lateral surface of the skull and the root of the zygoma, inserting primarilly into the medial side of the coronoid process along its anterior, superior, and posterior borders.

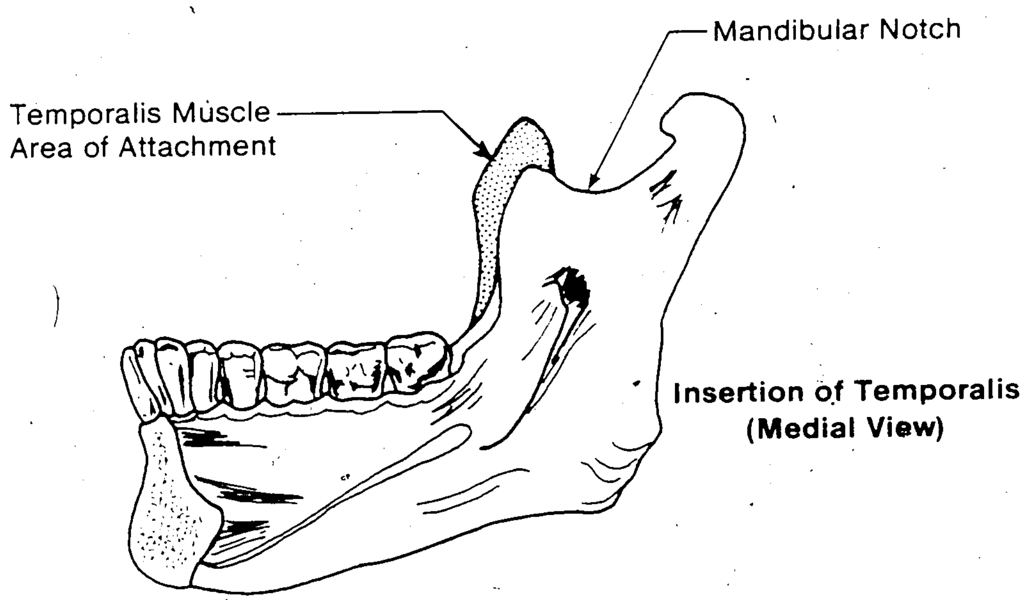
The deep and superficial masseter muscles both originate at the anterior portion of the zygomatic arch and insert on the lateral external surface of the mandibular ramus (figure 9). The superficial masseter muscle is lateral to the deep masseter muscle, and originates anterior to the deep masseter muscle.

The medial (internal) pterygoid muscle originates from the pterygoid plate of the sphenoid bone and inserts on the medial side of the mandibular angle and ramus, rising superiorly to the mandibular foramen (figure 10). This muscle is a counterpart of the masseter muscles and its direction in the sagittal plane is somewhat between the deep and superficial masseter muscles.

The lateral (external) pterygoid is composed of two heads. The larger inferior head has its origin at the

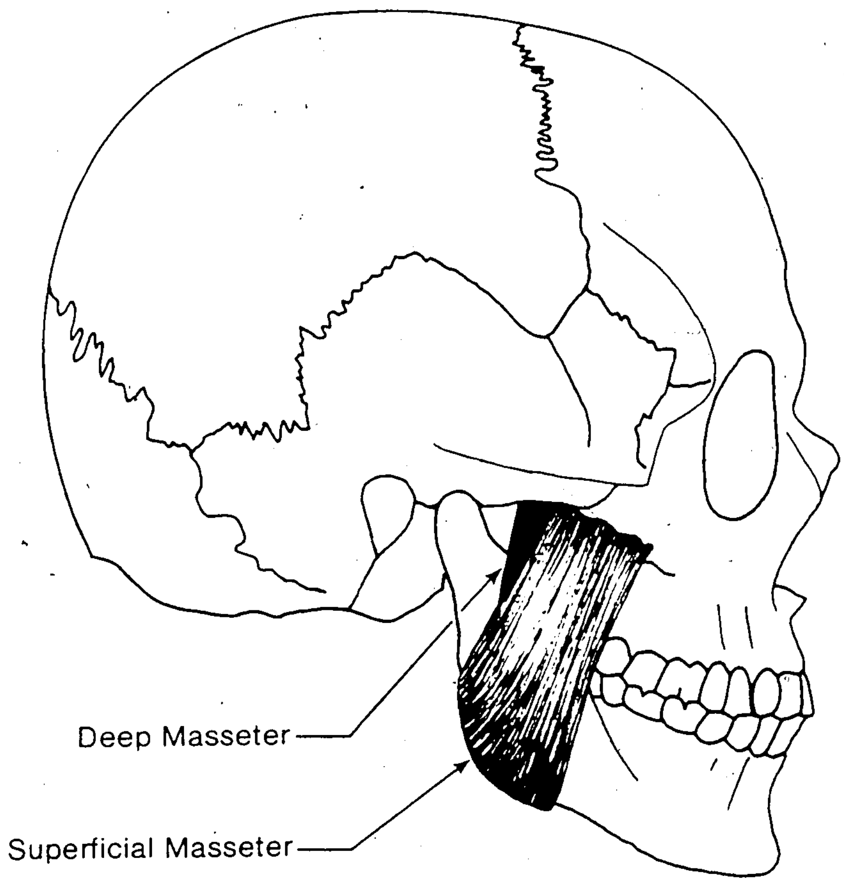


Temporalis Muscle (Lateral View)



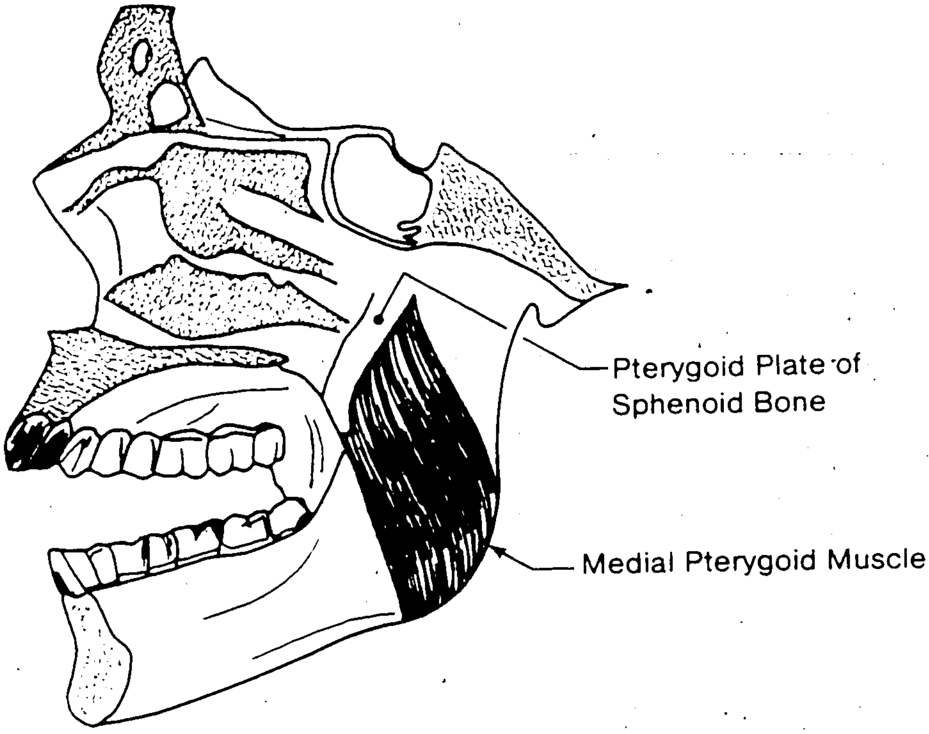
Temporalis Muscle

FIGURE 8

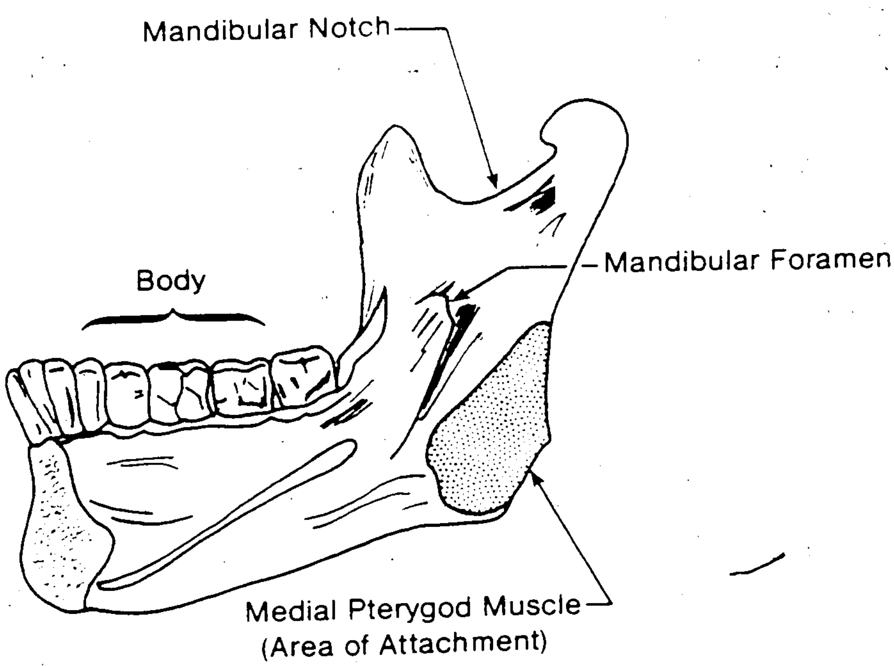


Masseter Muscle

FIGURE 9



Medial Pterygoid Muscle – Medial View



Medial Pterygoid Muscle Insertion – Medial View

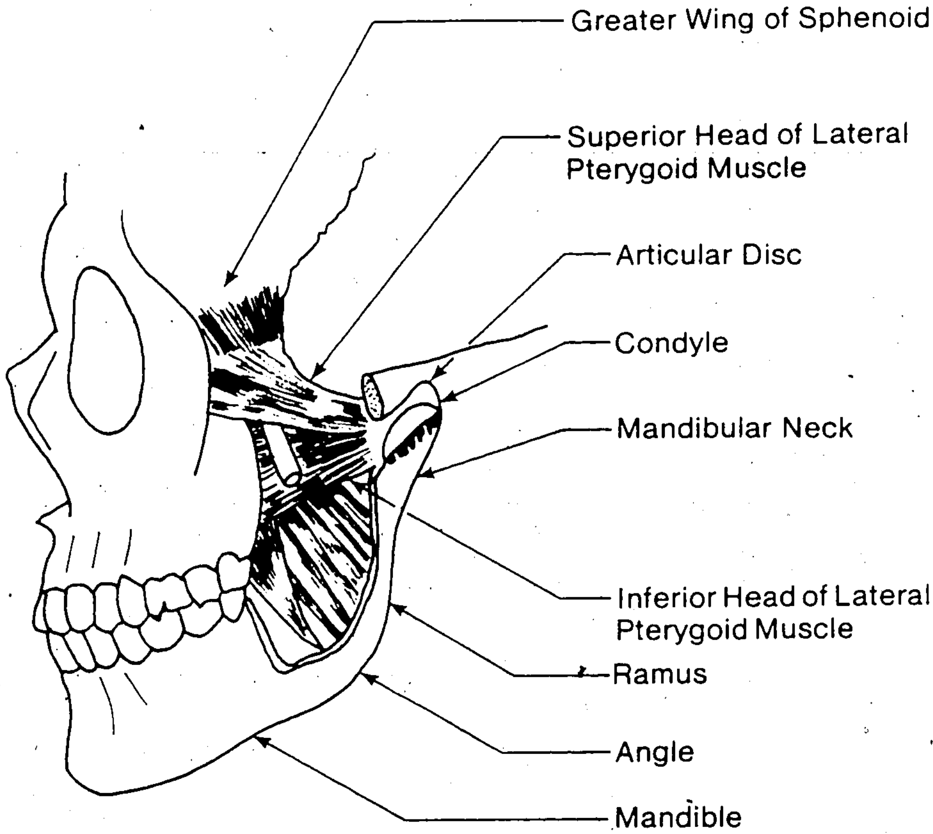
Medial Pterygoid Muscle

FIGURE 10

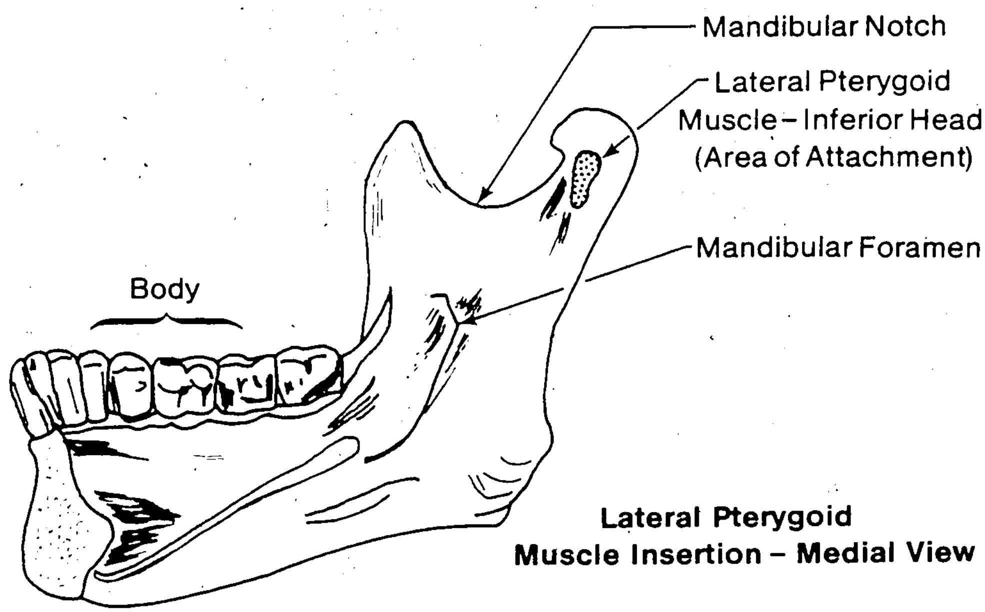
lateral pterygoid plate, while the superior head originates from the greater spenoid wing. The superior head attaches to the anterior portion of the articular disc and condyle and the inferior head inserts into the medial and anterior surfaces of the neck and condylar process (figure 11). During the power stroke, the inferior head is inactive, while the superior head exerts a stabilizing force, helping to hold the condyle against the articular eminence [46].

1.2 LITERATURE REVIEW

For many years researchers have debated as to whether the mandible functions as a lever or link. By functioning as a lever, it is meant that the TMJ functions as the fulcrum of a lever system, in which the condyles are loaded; a link refers to a case where the condyles are off-loaded during forceful occlusal contact. Due to the variability in muscle forces and directions, both possibilities could exist. At the heart of this debate was the issue of TMJ loading: could the joint support a load? Proponents of the link theory presented a seemingly convincing argument that the joint was poorly suited to support loads (based primarily on the fact that the roof of the glenoid fossa was extremely thin) [71, 78, 82], but recent research has shown that this is not correct (it is the eminence which supports the load, not the fossa) [30]. It is now generally accepted that the joint is capable of supporting substantial loads and is functionally loaded [6, 7, 15, 30, 66, 75].



Lateral Pterygoid Muscle – Lateral View



Lateral Pterygoid Muscle Insertion – Medial View

Lateral Pterygoid Muscle

FIGURE 11

Recent experiments by Hylander [31, 33, 34] using monkeys (macaques) have further substantiated the lever theory. Macaques were chosen since their masticatory system is somewhat similar to mans'. By implanting strain gauges *in vivo* on the neck of the mandible, it was found that the condyles were loaded during mastication and that for unilateral occlusion in the molar region the contralateral condyle supported more of the load. It is suggested that this is also the case in man.

Much work has been done to measure occlusal forces. Most researchers have found that occlusal forces increase as the point of occlusion moves from the incisors distally, although some researchers have found that forces on the second molar (M2) are sometimes less than those on the first molar (M1) [22, 23, 24, 43, 44, 58, 60].

Electromyography (EMG) has provided researchers with a better understanding of muscle activity. By inserting electrodes into the muscle fibers, the electrical activity of the muscle can be monitored. The greater the electrical activity of the muscle, the greater the force the muscle is exerting [37]. The masseter and temporalis muscles have been studied in much detail, but studies on the pterygoid muscles are few due to their inaccessibility. While EMG is inaccurate at predicting the exact force a muscle is exerting [67], it is a good predictor of when a muscle is active and can give some measure of the degree of activity. One important finding is that the anterior and posterior fibers of the temporalis muscle function independently so that the direction of the force of this muscle and its magnitude can vary considerably with the point of occlusal contact [2; 3].

Substantial research has been done in the area of joint remodelling and degeneration. There are several ways the TMJ remodels itself in response to increased joint loads: the shape of the condyle changes through bone deposition and resorbtion to distribute the load over a larger area in order to reduce stress levels; the thickness of the fibrous connective tissues increase; and the qualities of these tissues improve (resiliancy is increased) [47].

As the functional loads exceed the adaptive capabilities of the articular osseous and soft tissue, degeneration begins to occur. The thickness of the fibrous connective tissues decrease, their properties deteriorate, and much bone resorbtion occurs [54]. Failure to adapt to these excessive loads has been shown to lead to degenerative joint disease (osteoarthritis) [25, 36, 64].

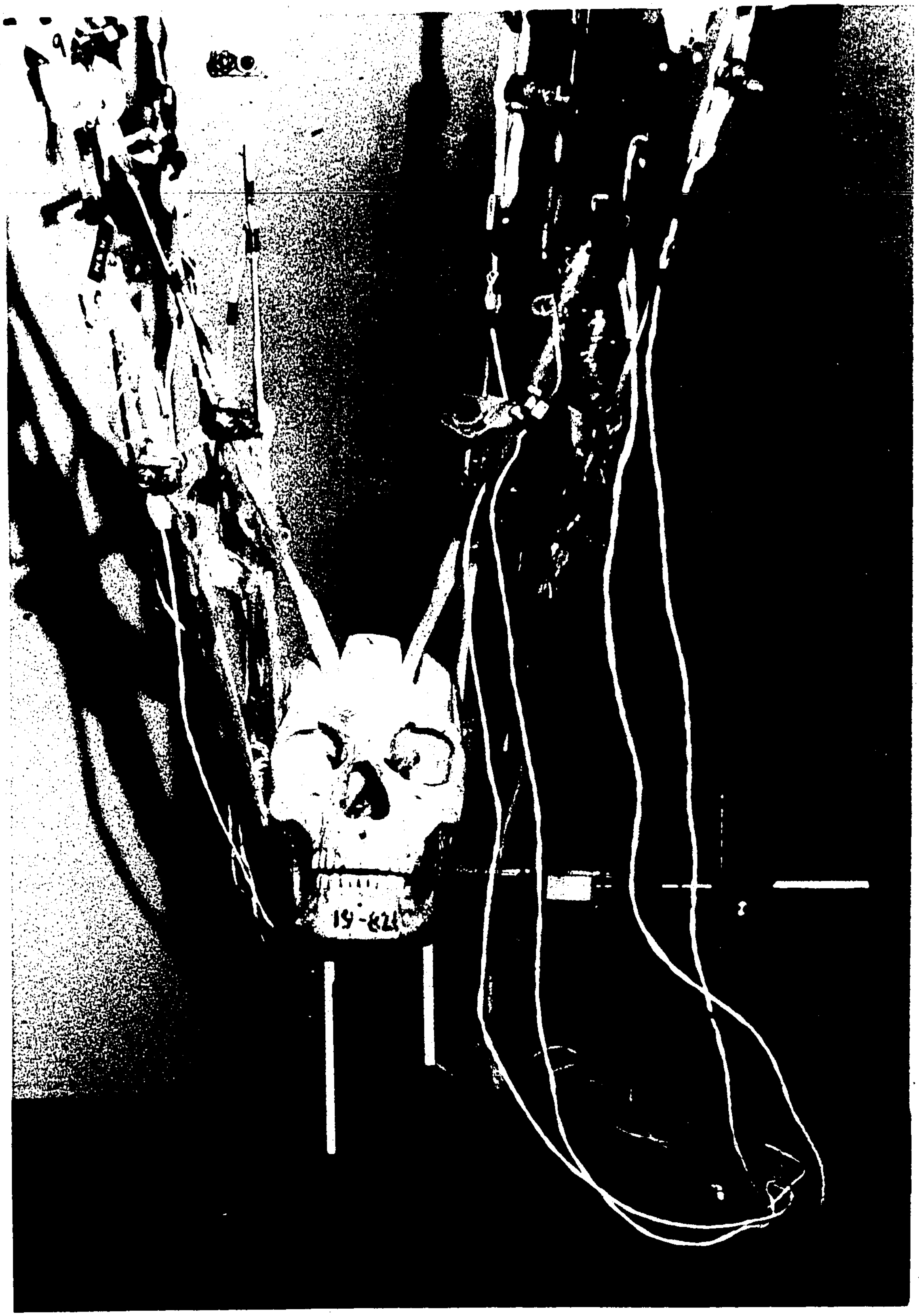
2. PHYSICAL AND MATHEMATICAL MODELS

Two models were developed: an *in vitro* model and a mathematical model. The *in vitro* model was developed from modifying an existing human skull while the mathematical model was based upon measurements from the *in vitro* model, allowing comparisons to be made. The *in vitro* model is described in section 2.1, and the mathematical model in section 2.2.

The *in vitro* model was developed to aid in conceptualization of the masticatory system and to make development of the mathematical model easier. It ensured accurate osseous anatomical proportions and allowed for modelling muscles in locations and directions as close to the *in vivo* situation as possible. Results from the *in vitro* model could then be compared with the mathematical model to check the accuracy of the mathematical model. The mathematical model allowed tests to be performed over a wide range of occlusal angles, occlusal positions, and muscle forces with much greater ease and accuracy than the *in vitro*.

2.1 IN VITRO MODEL

The *in vitro* model utilized a skull with Angles' Class I occlusion [51]. The skull was attached to a test frame and "muscle" forces were applied to the mandible (figure 12). Unilateral occlusal loads were then measured together with the resulting bilateral TMJ loads.



IN VITRO TEST ARRANGEMENT
FIGURE 12

2.1.1 SKULL MODIFICATIONS - PHYSICAL ARRANGEMENT

In order to hold the skull securely, it was attached to the test frame through the foramen magnum. The base of the posterior cranial fossa was filled with epoxy[3] along with the exterior surface of the occipital bone surrounding the foramen magnum to create two flat and parallel surfaces. A hole was drilled through the epoxy and foramen magnum, allowing the skull to be bolted to the test frame (figure 13). To prevent the possibility of fracturing the facial bones, the maxilla was reinforced with a steel plate anchored to the foramen magnum support (figure 14).

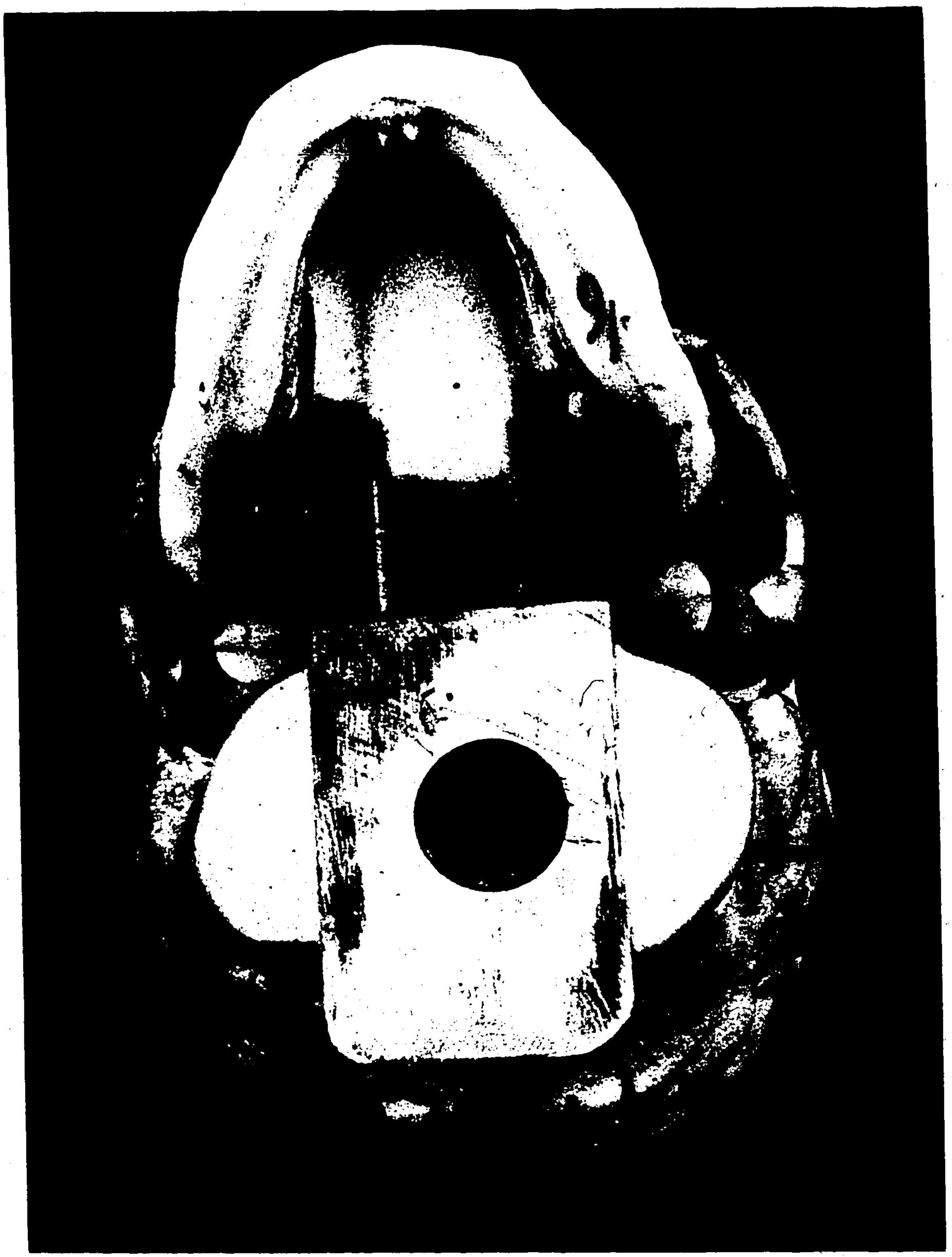
Since the test was static, the articular disc was modelled simply by using a material which would give a close approximation of the disc thickness between the condyle and eminence. This was facilitated by using Reprosil Putty[4] to fill the space between the condyles and their fosse and eminences while maintaining full occlusal contact in the molar region.

As occlusal loads are greatest when the bite opening is small [3], it was decided to keep the bite opening at a minimum while measuring occlusal loads. Since the smallest practical design for a bite force transducer (BFT) was 8.9 mm thick, the molar section of both the maxilla and the mandible were modified. The molar teeth were encased

----

[3] Araldite 509 combined with micro glass spheres - CIBA-Geigy Canada Limited.
[4] Reprosil Putty, DeTrey AG.

SKULL MODIFICATION – TOP VIEW
FIGURE 13



SKULL MODIFICATION – BOTTOM VIEW
FIGURE 14

in quartz resin[5] and then milled on a vertical milling machine to obtain two flat and parallel surfaces separated on both the left and right sides by 8.9 mm with a bite opening of 2.0 mm (bite opening measured at the central incisors) (figure 15). This allowed the BFT to be easily placed anywhere in the molar region to measure occlusal loads at a small bite opening. The removal of the cusps of the molar teeth was also desirable to enable contact over the full measuring surface of the BFT to avoid errors in occlusal force measurements. The occlusal angle thus formed was 90 degrees (angle between coronal plane and the line of action of the occlusal force as viewed in the axial plane) (figure 16).

The BFT was fabricated from tool steel and consisted of two beams clamped at one end and held apart by knife edges at the other end and at the centre (figures 17 and 18). Brass plates (slightly narrower than the bite table of the *in vitro* model) were bonded to the beams to ensure that the occlusal load would always be applied to the same place on the transducer. Strain gauges were mounted on the outer surface of the thinner section of the beams, so that the transducer would be very sensitive to small loads. These gauges were both in tension when the bite force was applied. Each gauge was connected in a Wheatstone quarter bridge circuit (figure 19).

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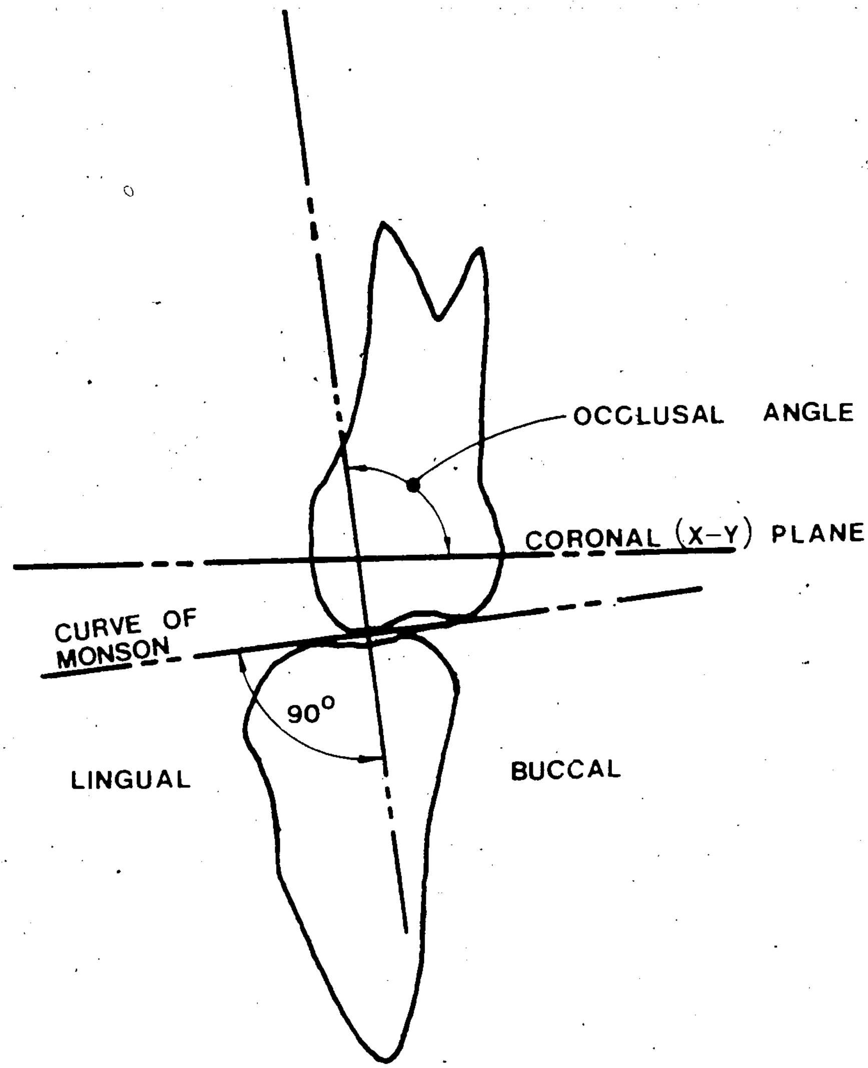
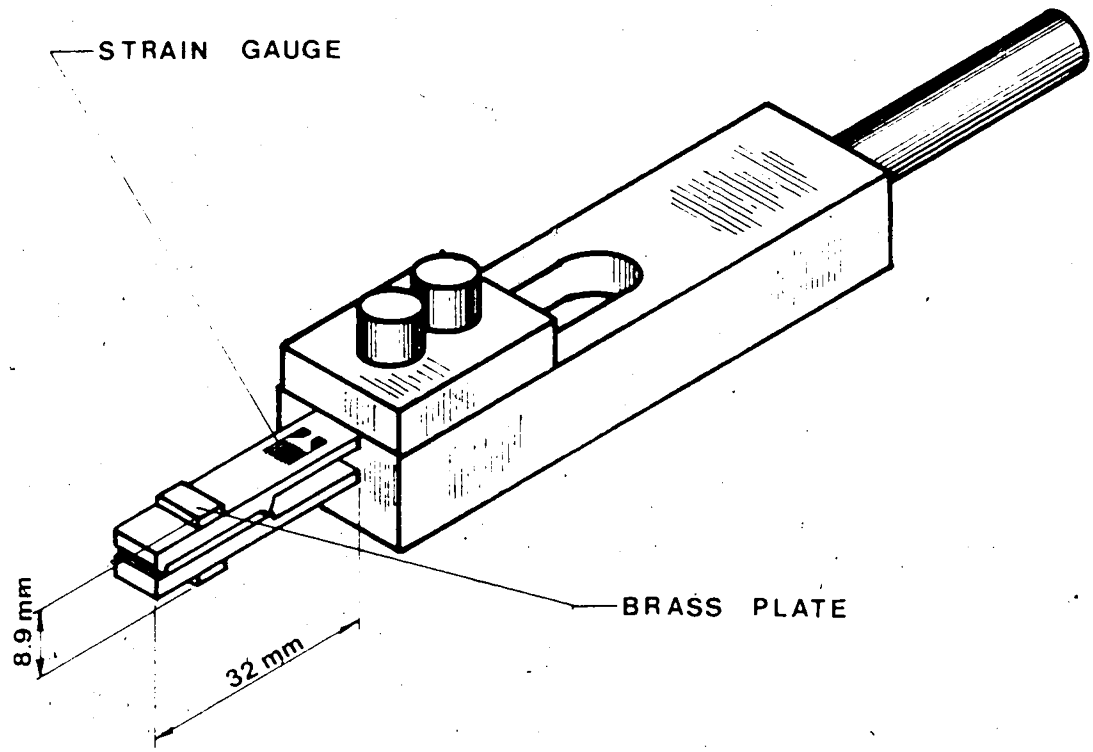
[5]Concise Composite, 3M Dental Products.

MOLAR TEETH ALTERATION
FIGURE 15



OCCLUSAL ANGLE

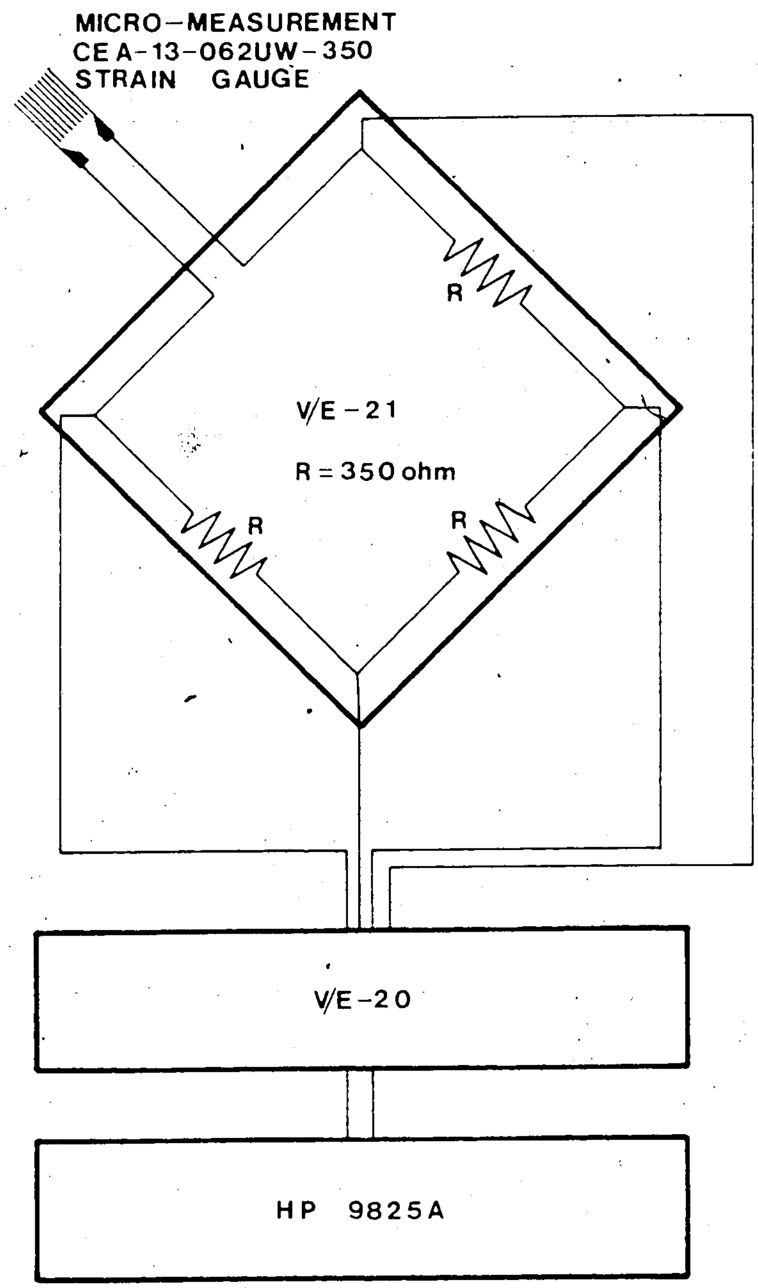
FIGURE 16



BITE FORCE TRANSDUCER

FIGURE 17

BITE FORCE TRANSDUCER
FIGURE 18



QUARTER BRIDGE CIRCUIT

FIGURE 19

The muscles of mastication were modelled using kevlar[6]. The muscles modelled were: the deep masseter, the superficial masseter, the medial pterygoïd, the anterior temporalis and the posterior temporalis. The lateral pterygoid was not included, as the force exerted by this muscle was believed to be comparitively small and it posed great difficulties to include it in the model (refer to section 3.3 for an analysis of this muscles' significance). The origin and insertion of the muscles were determined by interpreting osseous irregulartities produced by the original muscle attachments and were refined with reference to several anatomy texts [12, 29, 68, 72]. The kevlar strands were bonded to the mandible with epoxy[7] along the area of muscle insertion (figure 20). The area of insertion was first etched with phosphoric acid, rinsed with distilled water, and then air dried to obtain a surface suitable for the application of the epoxy. The "muscles" were aligned by passing the kevlar through the area of origin to a muscle force transducer (MFT). The medial pterygoid muscles required removal of bone in the anterior cranial fossa lateral to the crista galli and sections lateral to the midline of the frontal bone in order to pass the kevlar through the skull (figure 21). The zygomatic arch was also partially removed in order to align the masseter muscles correctly (figure 22). Care was taken not to modify any anatomical parts critical to subsequent testing.

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[6] Uniaxial kevlar - Hallcraft Plastics Limited.
[7] Araldite 502, CIBA-Geigy Canada Limited.

KEVLAR ATTACHMENT

FIGURE 20

MODIFICATIONS FOR MEDIAL PTERYGOID
FIGURE 21

MODIFICATION FOR MASSETER MUSCLE
FIGURE 22

The kevlar strands were attached to the MFT's, which were attached to turnbuckles supported from the test frame. "Muscle" forces were applied by tightening the kevlar with the turnbuckle and the forces were measured by the MFT's. These transducers were fabricated from aluminium (figures 23 and 24). The offset centre section of each transducer caused one side to be in tension and the other side in compression due to the bending moment involved. By mounting one strain gauge on the tension side and one on the compression side, a temperature compensated half bridge circuit was formed (figure 25).

Measuring the TMJ loads was difficult for two reasons: the size of the joint was small and the force had to be measured in three orthogonal planes. This was done by utilizing two strain gauge instrumented cantilevered beams of aluminum, one for each joint (figure 26). Each beam was centered superior to the glenoid fossa, followed by bonding to the superior surface with epoxy[a]. The fossa and eminence were then cut away from the rest of the skull, so that any force applied by the condyle would by supported by the beam alone (figures 27 and 28). Each beam was instrumented with eight strain gauges; each gauge was connected in a quarter bridge configuration (figure 19). By analysing the output of each gauge, the direction and magnitude of the condylar load could be determined.

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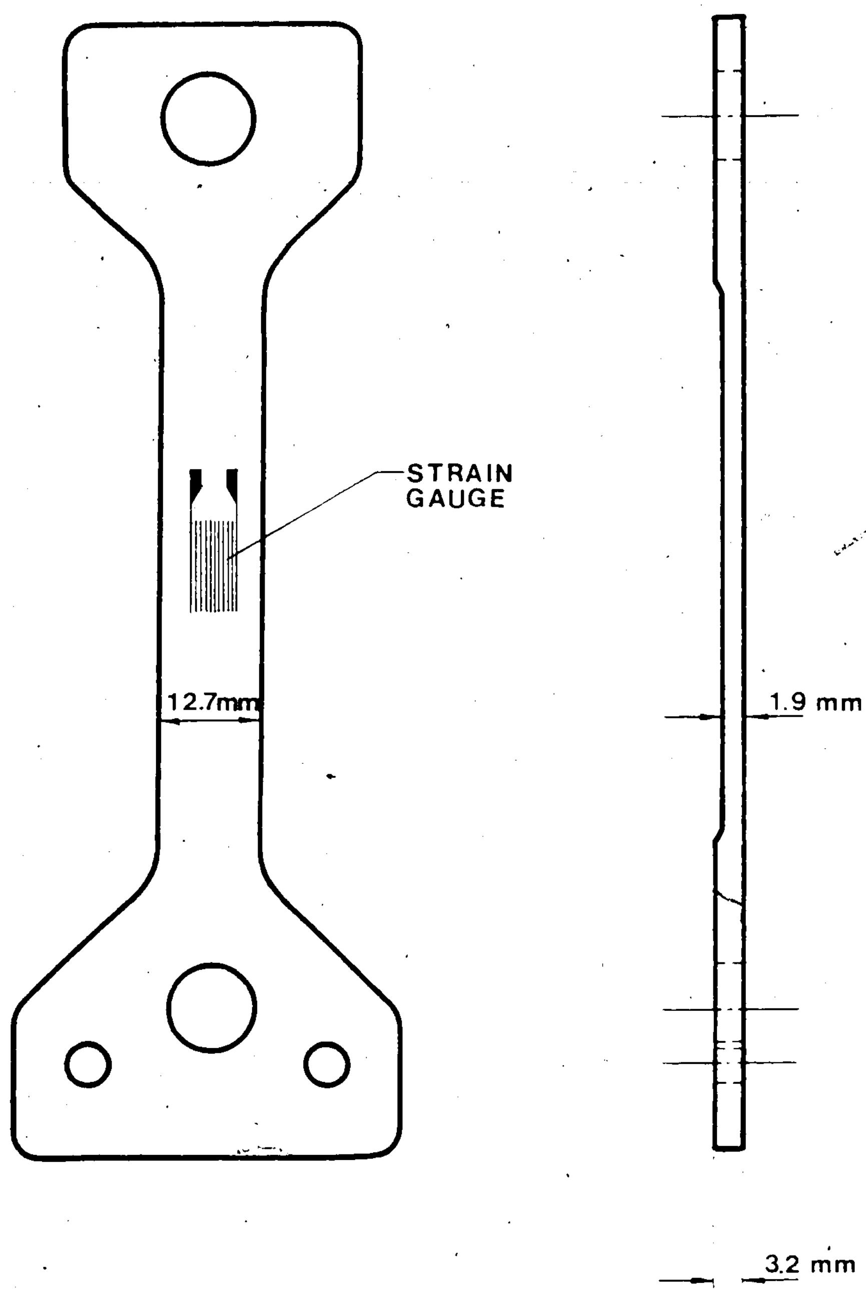
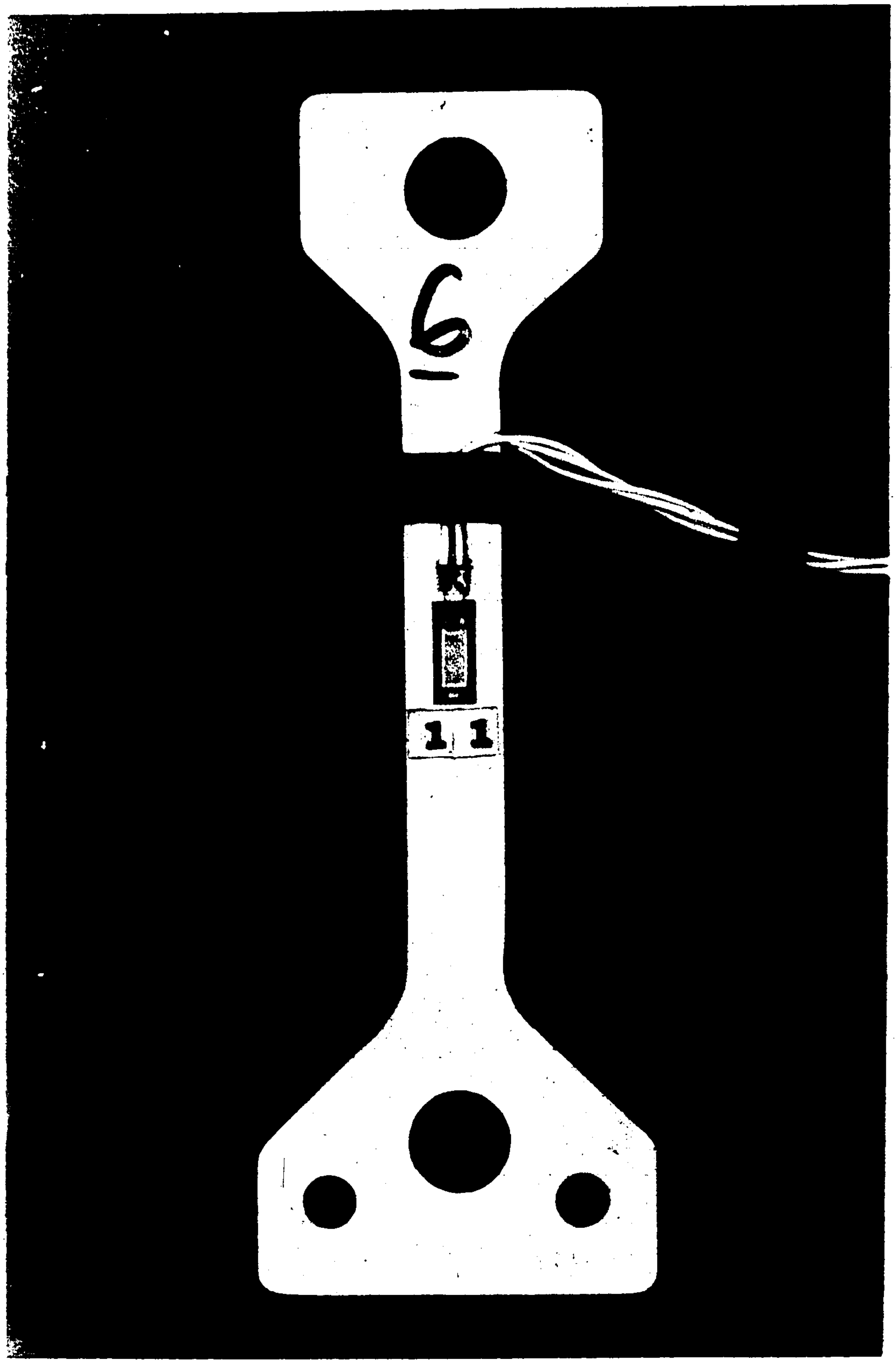
[a]Araldite 502, CIBA-Geigy Canada Limited.

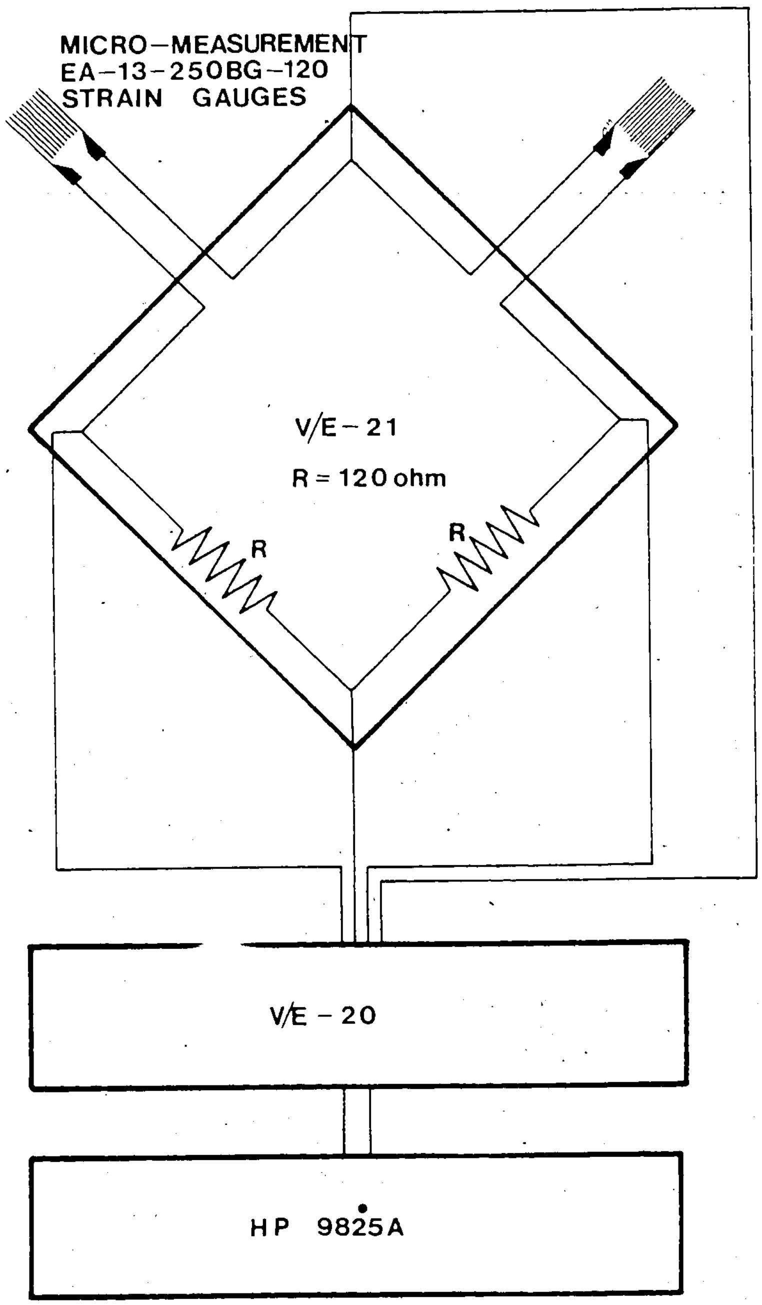


MUSCLE FORCE TRANSDUCER

FIGURE 23

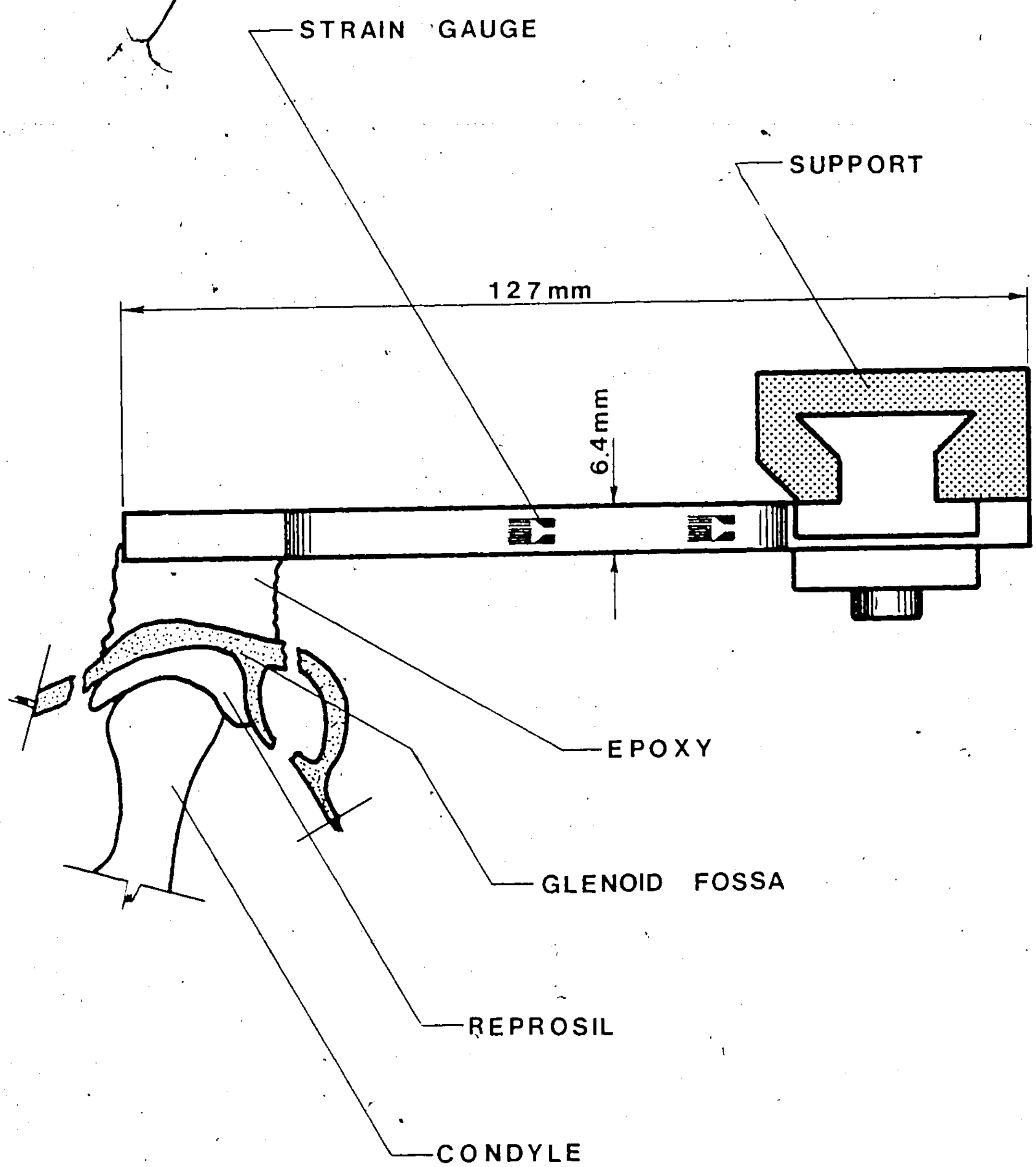


MUSCLE FORCE TRANSDUCER
FIGURE 24



HALF BRIDGE CIRCUIT

FIGURE 25



CONDYLAR FORCE TRANSDUCER

FIGURE 26

CONDYLAR FORCE TRANSDUCER
FIGURE 27

MODIFICATION FOR CFT
FIGURE 28

Each transducer was calibrated to determine strain output as a function of force. Refer to Appendix A for a detailed description of this calibration. All transducers were connected to a Vishay/Ellis Digital Strain Indicator[9] equipped for automatic scanning. Output from the Vishay/Ellis was read by a Hewlett Packard calculator[10] programmed to convert the strains into the applied and resultant forces.

Although the MFTs were temperature compensated, the BFT and CFTs were not. Potential problems stemming from the non-compensated bridge circuits were avoided by making use of the Vishay Ellis 20's capability to scan from bridge to bridge. Thus each individual bridge was excited only for a short period during data acquisition (less than two seconds per bridge), so that errors due to bridge imbalance caused by resistance heating of the strain gauges were not a significant factor. Furthermore, as all tests were done at room temperature (constant over the test period), bridge imbalance due to changing ambient temperature was avoided.

2.2 MATHEMATICAL MODEL

The mathematical model was based upon measurements taken from the skull used in the *in vitro* model. Force vectors were used to represent the muscle forces and vector analysis determined the resulting occlusal and joint loads.

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[9]Vishay/Ellis 20 combined with Vishay/Ellis 21 and Vishay/Ellis 25.
[10]Hewlett Packard 9825A desktop calculator.

A computer program was used to carry out the analysis once muscle forces and directions were specified.

2.2.1 PHYSICAL DESCRIPTION

The areas of muscle insertion were carefully marked out on the mandible of the *in vitro* model as previously mentioned. They were subsequently measured along with the position of the molar teeth and condlyes using a machinists microscope[11] and a vertical vernier[12]. The centroid of the area of muscle attachment was then evaluated and taken as the point of application of the muscle force according to the method of Dostál and Andrews [11]. The angles at which the muscles attached (with respect to the sagittal and axial planes) were measured from the *in vitro* model with a cathetometer comparator[13]. The muscles considered were the temporalis (divided into posterior and anterior sections), the medial masseter, the superficial masseter, the medial pterygoid and the lateral pterygoid (inferior head) muscles.

A Cartesian coordinate system centered at the apex of the left condylar process was used (figure 29), with unit vectors i, j, and k in the x, y, and z directions respectively. The centroid of the area of muscle attachment was then expressed as a position vector in terms of i, j, and k. The muscle force vector was expressed in terms of

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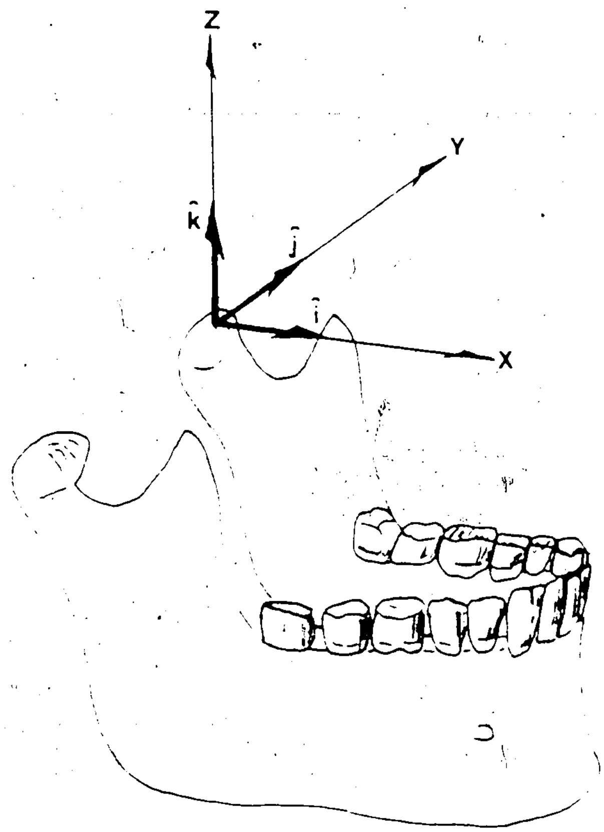
[11]Leitz Universal Toolmakers Microscope Model UWM.
[12]Brown and Sharpe Vernier Height Gauge.
[13]Precision Tool and Instrument Company Limited.



COORDINATE SYSTEM

FIGURE 29

its' magnitude and a normalized vector in the direction of the force. Table 1 lists the mucle force vectors and their position vectors. The model was made symmetric about the mid-sagittal plane, removing any minor asymmetries of the *in vitro* model.

After specifying the muscle forces and the angle of occlusal contact, vector analysis was used to solve for the condylar forces and directions , as well as the occlusal forces. The equations used were: the sum of the forces must be zero in the x, y, and z directions; and the sum of the moments about the x, y, and z axes must be zero. It was also assumed that the reaction on the left condyle would be the same as that upon the right condyle in the direction perpendicular to the sagittal plane (ie. in the y-direction). This assumption was necessary to reduce the problem to a statically determinate one. The error generated by this assumption was small since the reaction in the y-direction was small in comparison with the total joint load.

2.3 MUSCLE FORCES

One of the major unknowns in both models was the value of the various muscle forces in the configurations tested. Because of the uncertainty of these force values, muscle forces were determined from two parameters: muscle cross-sectional area and electromyographical (EMG) data. The *in vitro* model was tested with forces based on

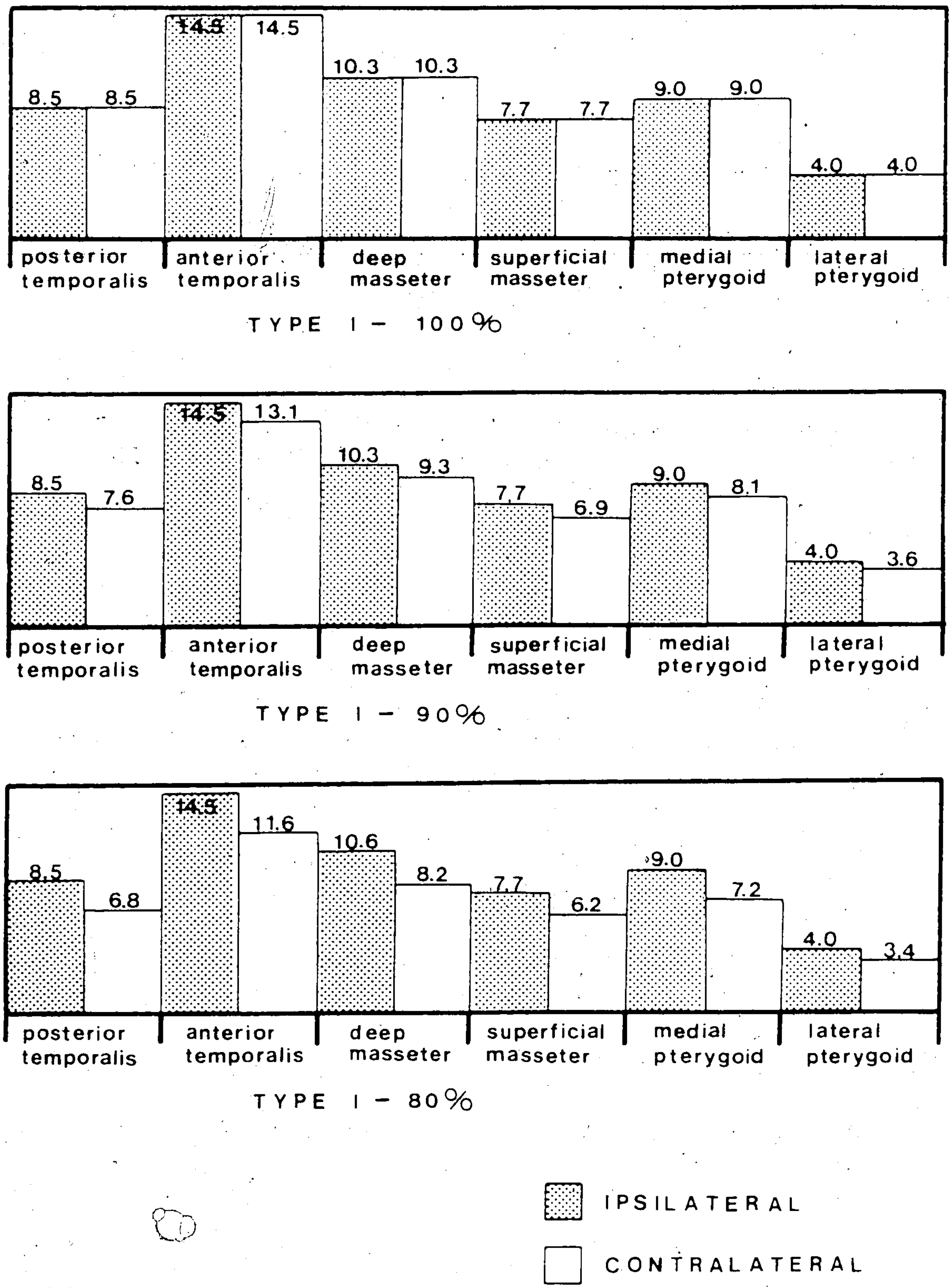
TABLE 1: MUSCLE PARAMETERS

MUSCLE	POSITION VECTOR	UNIT FORCE VECTOR
*****************	*****************	*****************
LEFT POST. TEMP.	1.35i-0.18j-0.00k	-0.76i+0.10j+0.64k
LEFT ANTR. TEMP.	1.48i-0.20j-0.26k	-0.34i-0.07j+0.94k
LEFT DEEP MASSR.	1.25i+0.00j-1.75k	-0.18i+0.27j+0.94k
LEFT SUP. MASSR.	1.25i-0.05j-1.75k	+0.15i+0.27j+0.95k
LEFT MED. PTERY.	0.88i-0.25j-1.75k	+0.03i-0.32j+0.94k
LEFT LAT. PTERY.	0.25i+0.00j-0.25k	+0.94i-0.25j-0.25k
RIGHT POST. TEMP.	1.35i-3.42j+0.00k	-0.76i-0.10j+0.64k
RIGHT ANTR. TEMP.	1.48i-3.40j-0.26k	-0.34i-0.07j+0.94k
RIGHT DEEP MASSR.	1.25i-3.60j-1.75k	-0.18i-0.27j+0.94k
RIGHT SUP. MASSR.	1.25i-3.65j-1.75k	+0.15i-0.27j+0.95k
RIGHT MED. PTERY.	0.88i-3.55j-1.75k	+0.03i+0.32j+0.94k
RIGHT LAT. PTERY.	0.25i-3.60j-0.25k	+0.94i+0.25j-0.25k

cross-sectional area only (Type I), while the mathematical model used both these forces (Type I) and forces based upon cross-sectional area and EMG data (Type II).

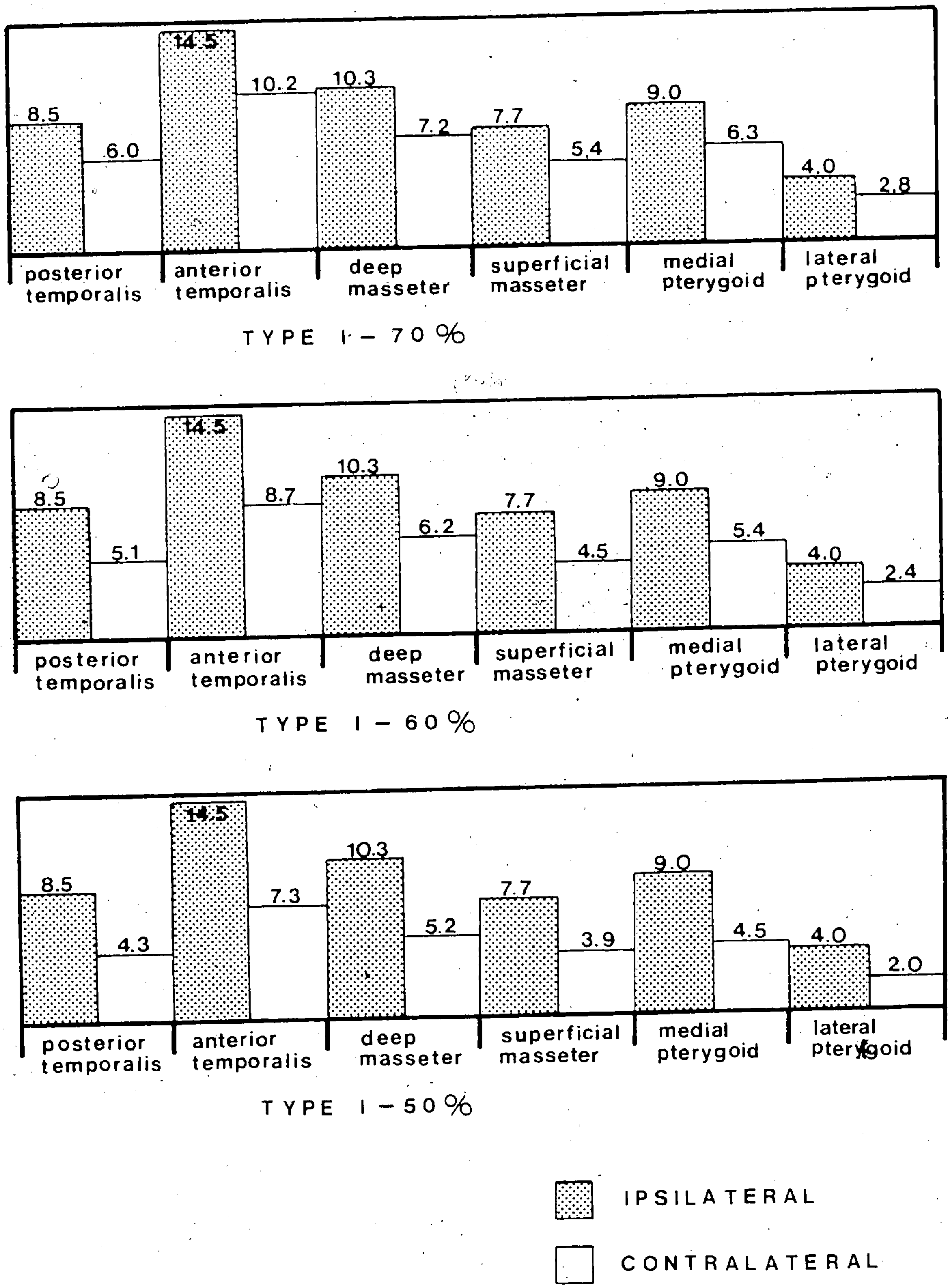
2.3.1 TYPE I

This scheme was based upon muscle cross-sectional area only. Muscle cross-sectional areas measured by Schumacher [16] were used. If a muscle had a cross-sectional area of ten square centimeters, it was given a force of ten units. The sum of all muscle forces was set to add to 100 units without the lateral pterygoid muscle force. With the lateral pterygoid muscle (superior head) force included, the muscle forces summed to 108 units. Research has indicated that the balancing side musculature is not as active as the working side during unilateral mastication [50], therefore the contribution of the balancing side muscles was decreased incrementally. The 100% case used identical muscle forces on the working and balancing sides. The 90% case used the same forces on the working side as in the 100% case, but only 90% of the forces were used on the balancing side. The 80% case used only 80% of the working side forces on the balancing side; the 70% case used only 70% on the balancing side; and so on. The contribution of the balancing side musculature was incrementally reduced by 10% each time, from 100% down to 50% (figures 30 and 31).



MUSCLE FORCES

FIGURE 30



MUSCLE FORCES

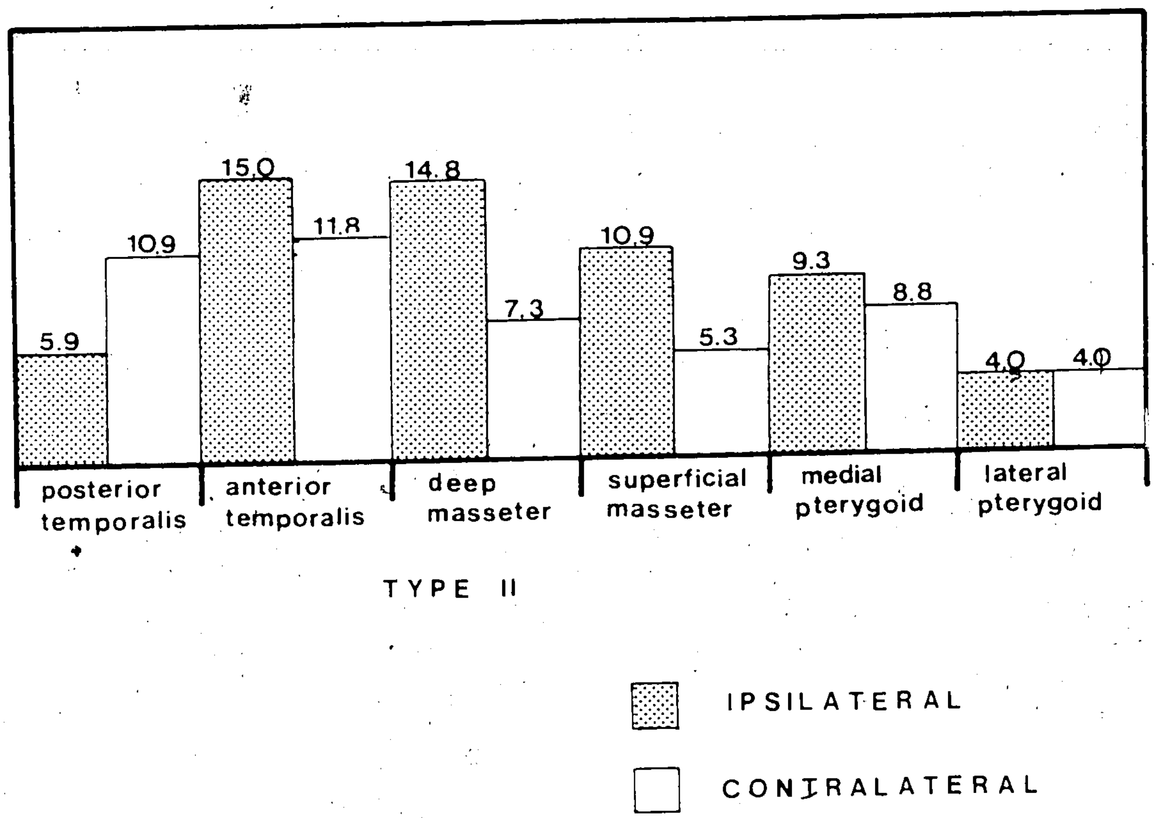
FIGURE 31

2.3.2 TYPE II

This scheme was based on cross-sectional area and EMG data. EMG data from Carlsoo, Pruim <u>et al</u>, Mushinoto <u>et al</u>, and Moller [8, 50, 52, 59] was used to form a basis of relative muscle activity. This was then used together with muscle cross-sectional a[illegible]etermine individual muscle forces. For example, if a [illegible]ad a cross-sectional area of 10 square centimete[illegible] given a maximum force capability of 10 units. If the EMG data indicated that it was 85% active, then the force exerted by this muscle was set at 8.5 units. As in Type I, the sum of all muscle forces was set to 100 units without the lateral pterygoid muscle force (figure 32). With the lateral pterygoid muscle (superior head) force included, the muscle forces summed to 108 units.

2.4 METHOD OF TESTING

Tests were performed on the *in vitro* model using Type I muscle forces (at 100%, 80%, and 60% levels, without inclusion of the lateral pterygoid muscle force). A minimum of three tests were performed at each occlusal positon at each force level. Each test was performed independently (ie. for each test, the BFT was repositioned, all bridges were balanced, and muscle forces were reapplied). Tests were also performed on the mathematical model using Type I muscle forces (without inclusion of the lateral pterygoid muscle



MUSCLE FORCES

FIGURE 32

forces, at 100%, 80%, and 60% levels) to compare results with the *in vitro* tests.

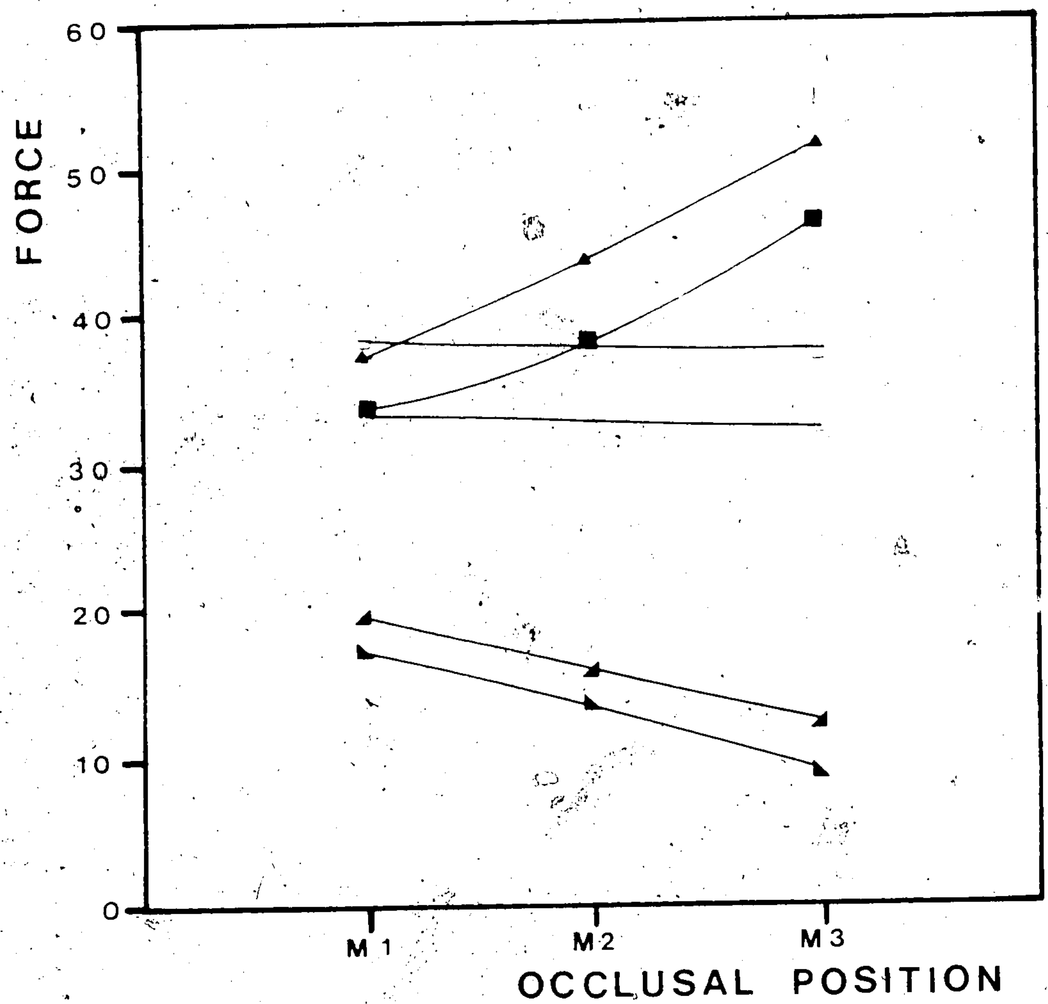
Further testing was done on the mathematical model using Type I (at 100%, 90%, 80%, 70%, 60% and 50% levels) and Type II muscle forces with the lateral pterygoid (superior head) muscle forces included and the occlusal angle (Figure 16) was varied from 75 degrees to 105 degrees in 5 degree increments. Tests to determine the resulting occlusal and TMJ loads to changes in muscle forces, muscle positions, and muscle angles were also performed on the mathematical model.

3. RESULTS AND DISCUSSION

3.1 MODEL COMPARISON

The *in vitro* model was tested with Type I muscle forces, without lateral pterygoid muscle force, at the 100%, 80%, and 60% levels, with the occlusal angle set to 90 degrees. Occlusal loads were measured at M1, M2, and M3 unilaterally. The mathematical model was also tested under the same conditions for comparison. These results are shown in figures 33, 34, and 35.

The results indicate that the models correlate reasonably well, as the same general trends are evident for occlusal and TMJ loads. The greatest deviation between test systems occurs in the ipsilateral TMJ loads for unilateral occlusion at M3 (30% for Type I muscle forces at the 100% level; 37% for Type I muscle forces at the 80% level; 34% for Type I muscle forces at the 60% level). Occlusal loads correlated much better; the largest difference was 19% (13% for Type I muscle forces at the 100% level at M2; 19% for Type I muscle forces at the 80% level at M3; 19% for Type I muscle forces at the 60% level at M2). The best correlation between models was found for contralateral TMJ loads, where the largest difference was 13% (13% for Type I muscle forces at the 100% level at M2; 5% for Type I muscle forces at the 80% level at M1; 6% for Type I muscle forces at the 60% level at M3).

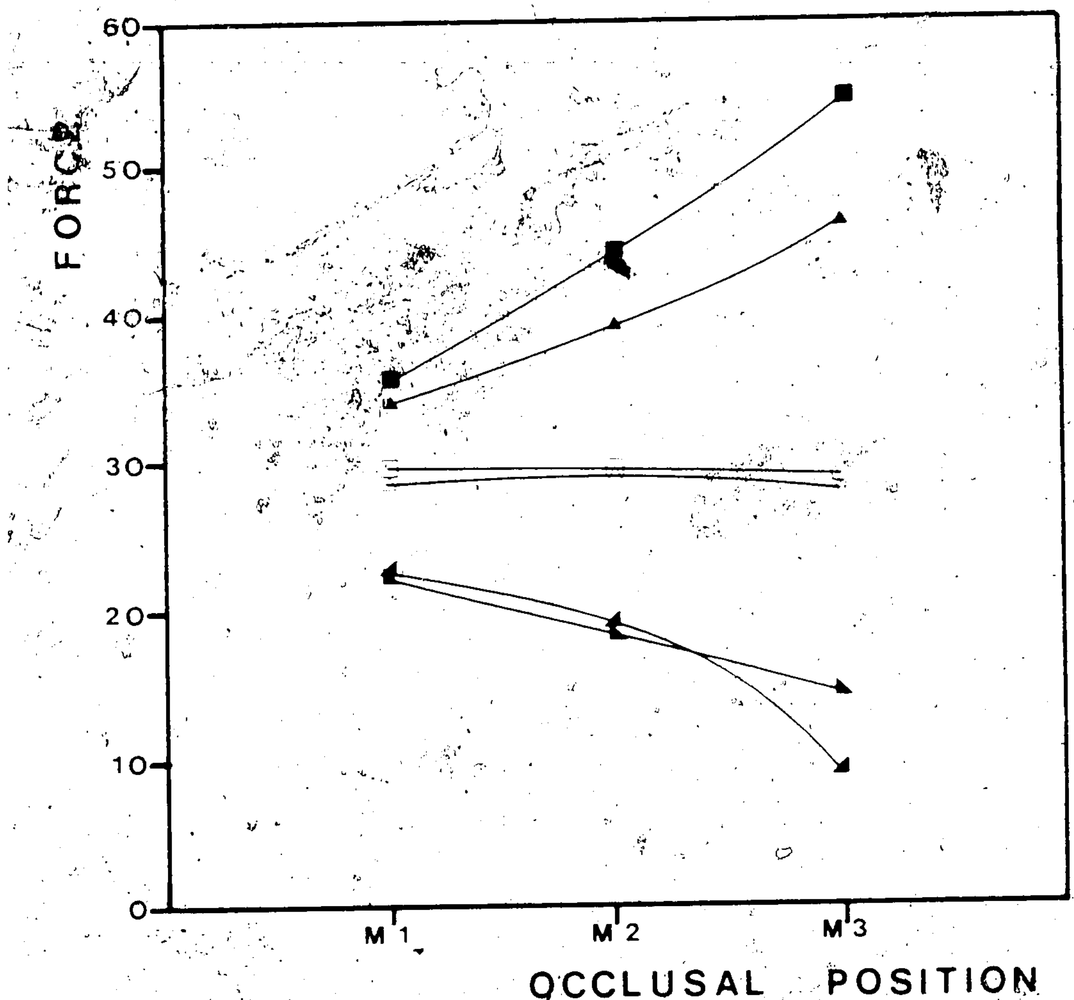


	OCCLUSAL LOAD	IPSI. TMJ LOAD	CONTRA. TMJ LOAD
IN VITRO	■	◢	
MATHEMATICAL	▲	◣	

TYPE I MUSCLE FORCES AT 100%

MODEL COMPARISON

FIGURE 33

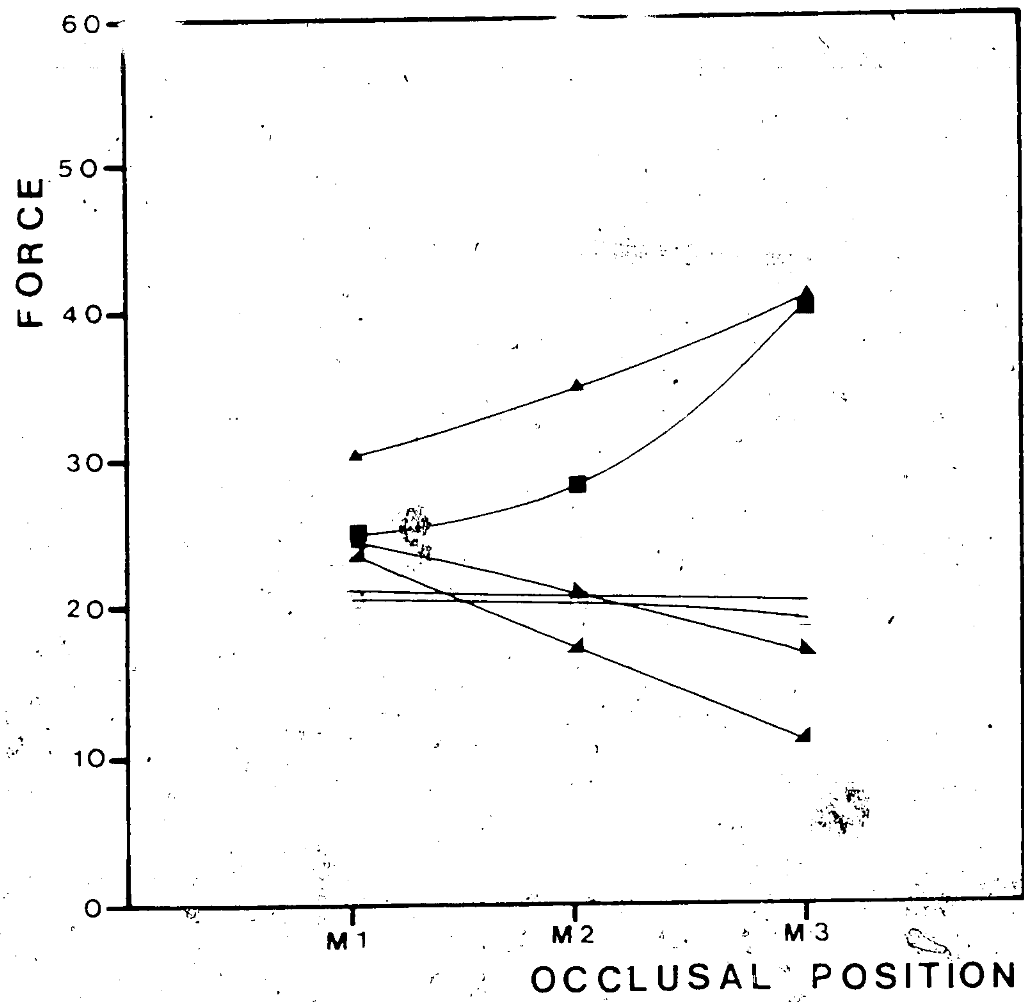


	OCCLUSAL LOAD	IPSI. TMJ LOAD	CONTRA. TMJ LOAD
IN VITRO	■	◢	
MATHEMATICAL	▲	◣	

TYPE I MUSCLE FORCES AT 80%

MODEL COMPARISON

FIGURE 34



	OCCLUSAL LOAD	IPSI. TMJ LOAD	CONTRA. TMJ LOAD
IN VITRO	■	◢	
MATHEMATICAL	▲	◣	

TYPE I MUSCLE FORCES AT 60%

MODEL COMPARISON

FIGURE 35

The deviation between the *in vitro* and mathematical models can be accounted for by several factors. First of all, due to the small forces which were to be measured, the force transducers were designed to be very sensitive. In the case of the MFTs, this also created sensitivity to torsional loads, which were often present from turnbuckle adjustments. Errors in measurement of muscle forces were estimated to be as large as twenty per cent.

The second factor causing discrepencies between models was the difference in muscle attachment points. The mathematical model used a single point (the centroid of the areaf the of insertion); the *in vitro* model used kevlar bonded over the full area of insertion. When attaching the kevlar strands from the mandible to the MFTs, it was impossible to ensure that all strands were evenly tightened. Thus, when muscle forces were applied, some strands were subjected to greater loads than others. The net effect of this was that the centroid of the applied muscle force no longer coincided with the centroid of the area of insertion, creating a discrepency between models.

Further inaccuracies between models resulted from the difficulty in positioning the kevlar "muscles" of the *in vitro* model. The angles at which the muscles were positioned were only within five degrees of the desired angle. This resulted from problems associated with bonding the kevlar to the mandible. Although great care was taken to attach the kevlar at the correct angle, this proved to be exceedingly difficult. Also, the kevlar strands exhibited a tendency to wick the epoxy away from the mandible before the epoxy cured, causing fibers to become hardened for up to 2 centimeters from the point of attachment. This created further problems, as the kevlar no longer had the flexibility required to properly adjust the attachment angle in this area.

The errors in force measurement in the mathematical model and differences between models in muscle attachment points and muscle attachment angles can be shown to account for the deviation in results of the models, as will be discussed further in the sensitivity analysis in section 3.7. It should be noted that each point plotted from the *in vitro* model represents the average of several tests (minimum of three). If one were to selectively disregard points which appeared erroneous, the *in vitro* model would correlate much better with the mathematical model.

3.2 JOINT LOADS

The two models illustrate that, for unilateral occlusion, the TMJ is loaded. This was found to be the case in every situation tested. Furthermore, contralateral TMJ loads will usually exceed ipsilateral TMJ loads. Figure 35 reveals that a point could be attained where the ipsilateral TMJ loads exceed contralateral TMJ loads. This occurs when the contralateral muscle forces have been reduced to 60% or less of the ipsilateral muscle forces. However, it is not likely that this would occur, as EMG data indicates that the contralateral muscle forces are generally in the order of 70 to 100 per cent as active as the ipsilateral side. Furthermore, as previously mentioned, studies by Hylander [33, 34] have shown the contralateral TMJ loads to be higher for monkeys with a fused symphisis (an animal with a somewhat similar masticatory system to the human system). Research by Caputo et al [66,75] has also indicated higher contralateral TMJ loads for unilateral occlusion.

Thus it can be said with some confidence that the TMJ is loaded, and contralateral TMJ loads will exceed ipsilateral TMJ loads for unilateral occlusion in the molar region. This illustrates a possible danger in losing teeth in the molar region: if, for example, molar teeth were lost unilaterally, the individual would be forced to masticate primarily on one side of the mandible only. The result of this would be that one TMJ (the contralateral) would always be subjected to higher loads than the other TMJ (the ipsilateral). Thus it is important to maintain occlusal surfaces in the molar region to avoid subjecting one TMJ to consistently higher loads than the other.

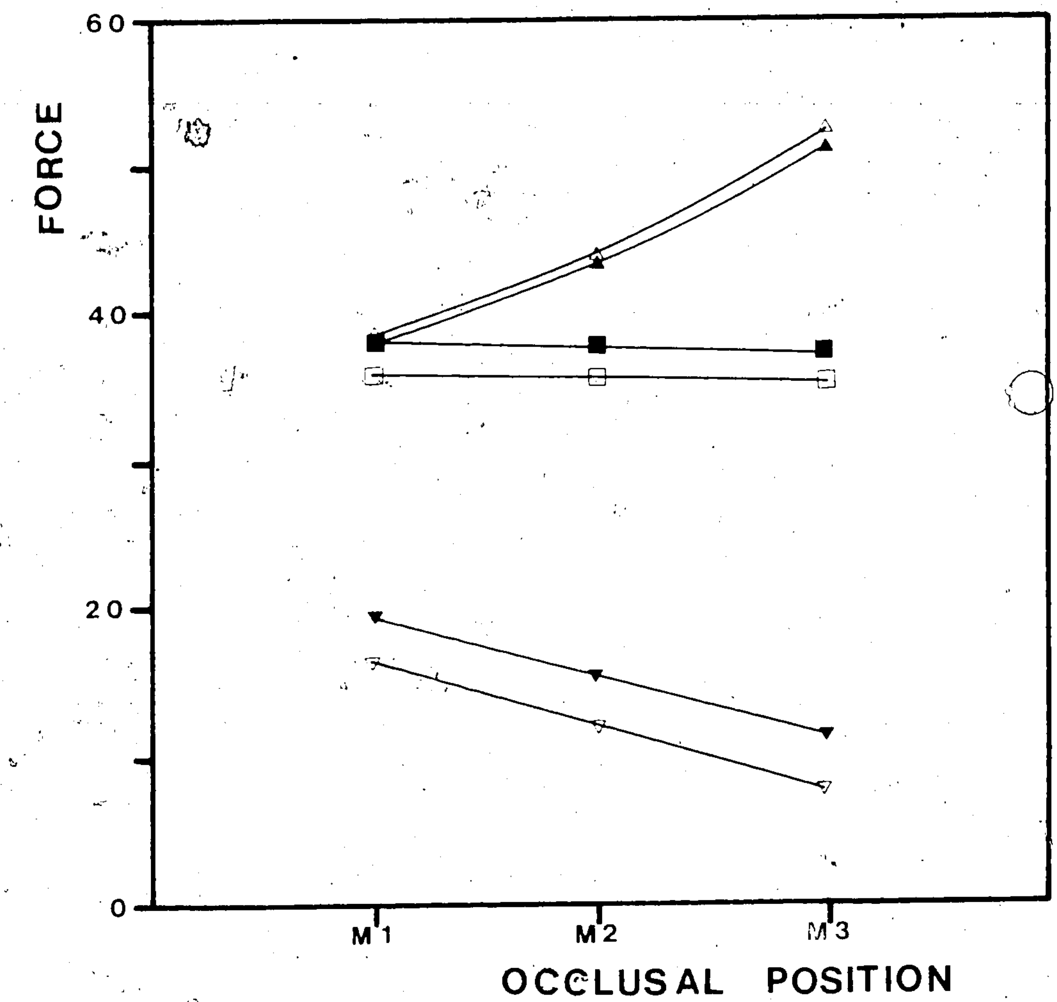
3.3 EFFECT OF LATERAL PTERYGOID MUSCLE FORCE

The mathematical model was tested with Type I muscle forces (at 100%, 90%, 80%, 70%, 60%, and 50% levels) both with and without lateral pterygoid muscle (superior head) force. Occlusal angles (Figure 16) were varied from 75 degrees to 105 degrees in 5 degree increments. Figure 36 compares results of these tests at the Type I - 100% level at an occlusal angle of 90 degrees. These results are typical for all cases tested, and show that, while lateral pterygoid muscle (superior head) force may slightly affect the magnitude of the TMJ loads, it will not change the trends involved. It was found that the major effect of this muscle was to change the angle of the TMJ reaction, which supports the contention of McNamara [46] that the superior head of the lateral pterygoid is primarily used for stabilizing the condyle during forcefull occlusal contact. The effect on occlusal loads was minimal, as one would expect from the point of insertion of this muscle.

3.4 EFFECT OF OCCLUSAL POSITION

Tests were performed with the mathematical model to evaluate the effect of occlusal position, using both Type I (at the 100%, 90%, 80%, 70%, 60%, and 50% levels) and Type II muscle forces. Occlusal position was tested at M1, M2 and M3 for occlusal angles of 75 to 105 degrees.

Figures 37, 38, and 39 illustrate the effect of occlusal position on occlusal force and TMJ reactions. As the occlusal position is moved from M1 to M3 (from the first molar distally), occlusal loads increase, ipsilateral TMJ loads decrease, and contralateral TMJ loads decrease slightly. Figures 37, 38, and 39 are results for an occlusal angle (Figure 16) of 90 degrees; similar trends were

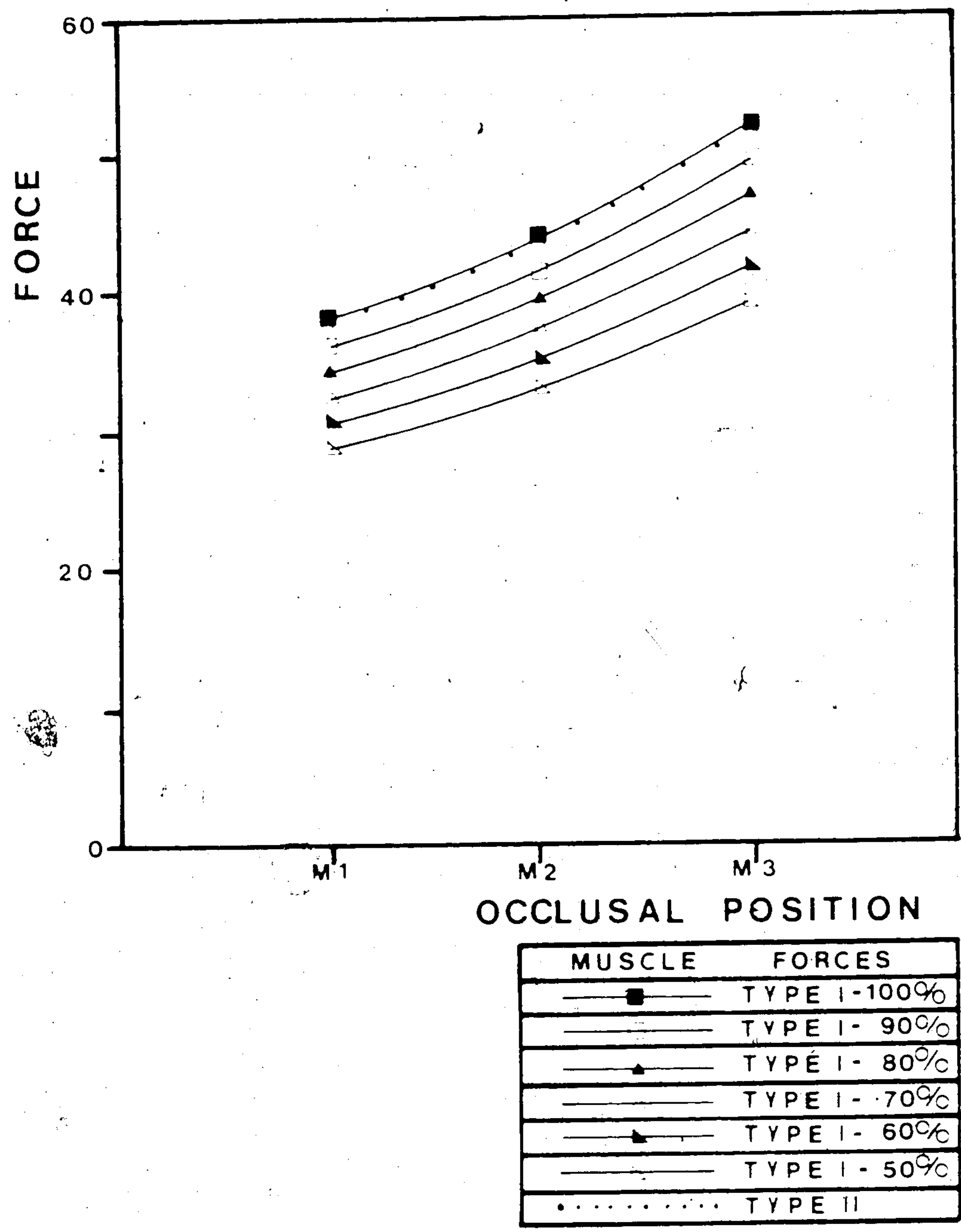


	OCCLUSAL LOAD	IPSI. TMJ LOAD	CONTRA. TMJ LOADS
WITHOUT LPM	▲	▼	■
WITH LPM	△	▽	□

TYPE I MUSCLE FORCES AT 100%
OCCLUSAL ANGLE AT 90°

EFFECT OF LATERAL PTERYGOID MUSCLE FORCE

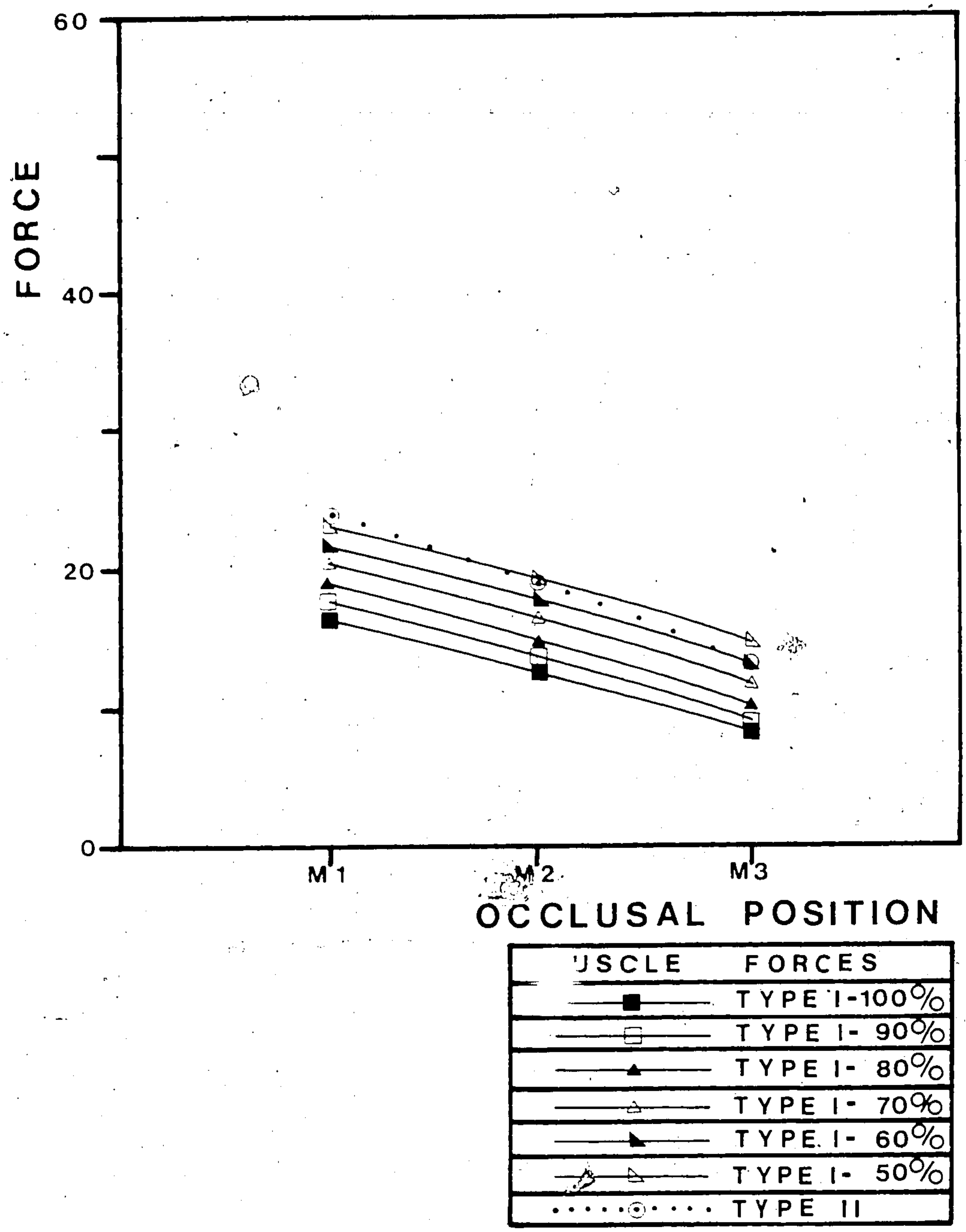
FIGURE 36



OCCLUSAL ANGLE AT 90°

EFFECT OF OCCLUSAL POSITION ON OCCLUSAL LOADS

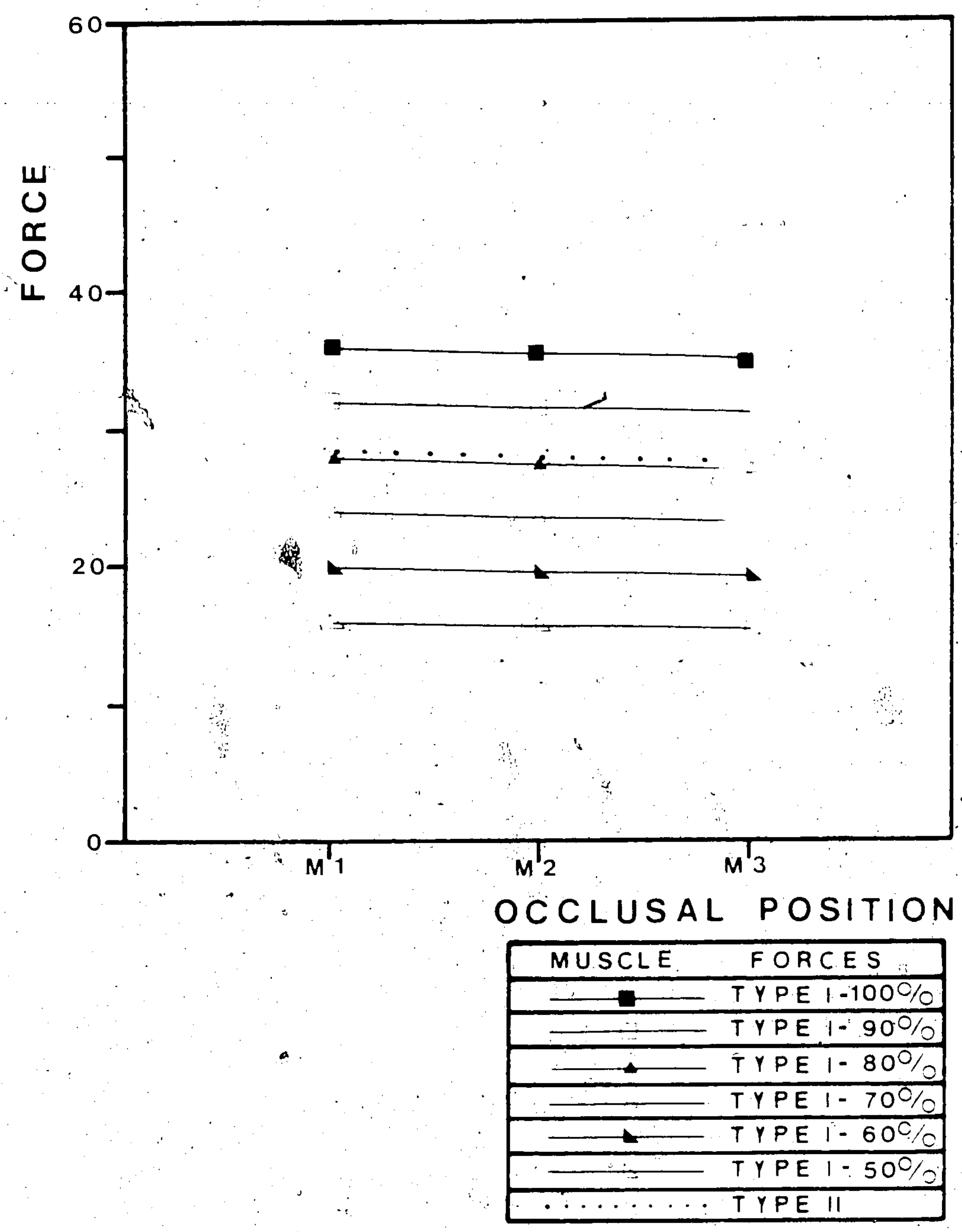
FIGURE 37



OCCLUSAL ANGLE AT 90°

EFFECT OF OCCLUSAL POSITION ON IPSILATERAL TMJ LOADS

FIGURE 38



OCCLUSAL ANGLE AT 90°

EFFECT OF OCCLUSAL POSITION ON CONTRALATERAL TMJ LOADS

FIGURE 39

observed for all occlusal angles examined (75 to 105 degrees).

As mentioned in section 1.2, some researchers have found that occlusal loads increase, while others have found that occlusal loads decrease, as the point of occlusal contact is moved distally. Both the *in vitro* model (with Type I muscle forces) and the mathematical model (with Type I and Type II muscle forces) support the finding of increased occlusal loads as the point of occlusal contact is moved in the distal direction.

However, there are two distinct areas in which both models differ from the *in vivo* situation. The first is that muscle forces were not adjusted for the three different points of occlusion. Since M1 erupts much sooner that the other molar teeth, it is entirely possible that the muscles of mastication may be selectively more developed for applying occlusal force at M1 in some individuals. This could account for the finding of decreased occlusal loads as the point of occlusion was moved from M1 distally.

The second difference between the *in vivo* situation and the models in this study was that the Curve of Spee was ignored in the models (ie. anteroposterior inclination of the occlusal plane was not changed from M1 to M3). Studies by Okane et al [55] have shown that bite force decreased when the occlusal plane was changed +/-5 degrees (in the anteroposterior sense) from the normal bite plane. Thus it is expected that the deviation of the occlusal plane from the Curve of Spee in the models should cause a decrease of occlusal force.

3.5 COMPARISON OF MUSCLE FORCES

Figures 37, 38, and 39 also show the effect utilizing the various Type I and Type II muscle forces. As the contralateral muscle force contribution is decreased (ie. from Type I at 100% to Type I at 50%), occlusal loads and contralateral TMJ loads decrease, while ipsilateral TMJ loads increase. As previously mentioned, this will cause a point to be reached where ipsilateral TMJ loads exceed contralateral TMJ loads.

Type II muscle forces should only be compared with Type I at 100%, as this is the only case where Type I muscle forces sum to the same total as Type II muscle forces. Figure 37 reveals that both schemes develop nearly identical occlusal loads (Type II occlusal loads are slightly less that Type I at 100%). Figure 38 shows that ipsilateral TMJ loads are greater for the Type II muscle forces. Figure 39 indicates that contralateral TMJ loads are less for the Type II muscle forces.

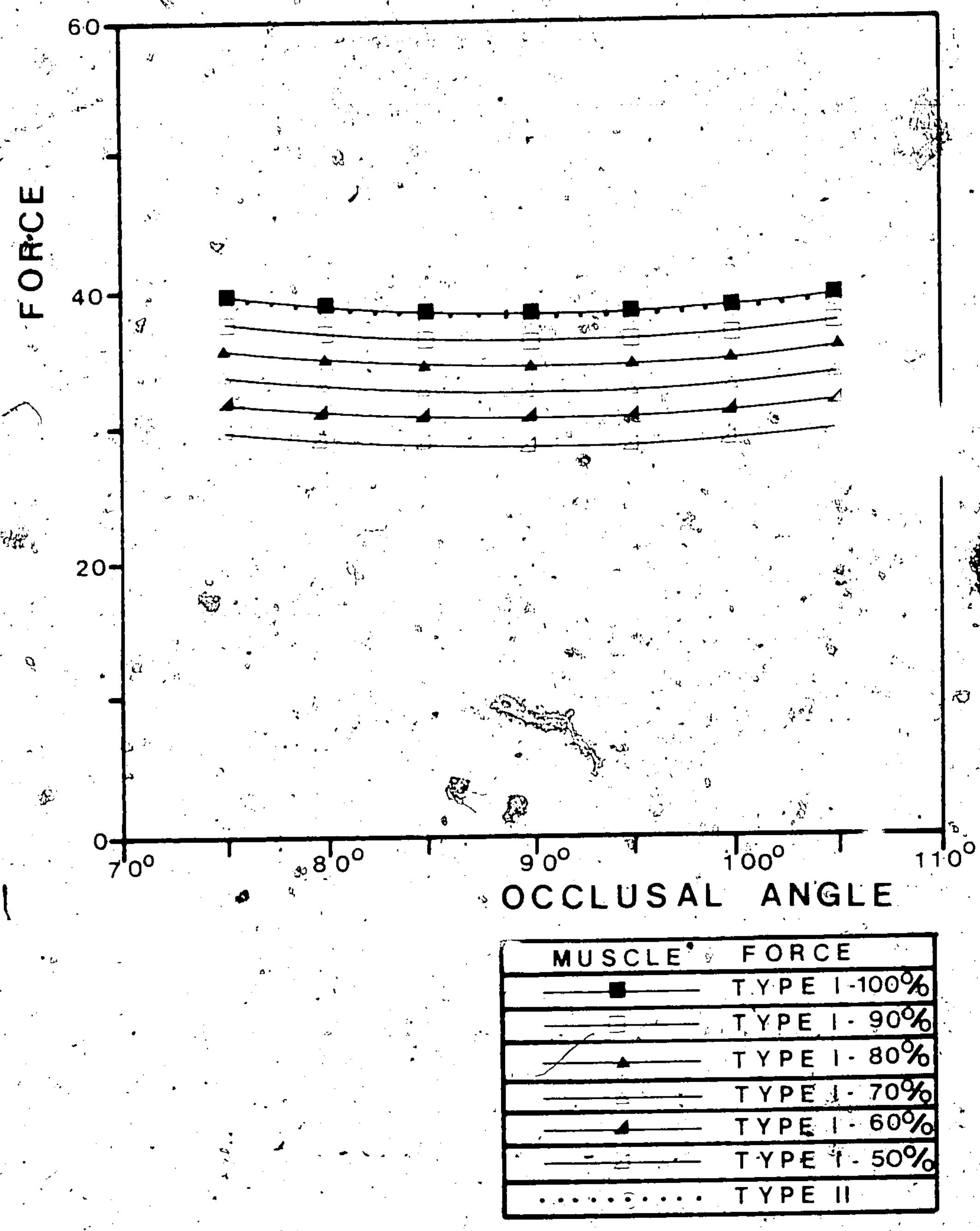
Since the peak TMJ loads are reduced (contralateral TMJ loads are decreased) without significantly altering the occlusal forces when Type II muscle forces are used instead of Type I muscle forces, the Type II muscle forces result in an improved situation from the point of view of TMJ loads. Although ipsilateral TMJ loads are increased, they are still below the contralateral TMJ loads. Thus is expected that the Type II muscle force scheme more closely models the *in vivo* situation (on the basis that the *in vivo* system would most certainly adapt itself to keep TMJ loads at a minimum without sacrificing occlusal force).

3.6 EFFECT OF OCCLUSAL ANGLE

The mathematical model was tested with both Type I and Type II muscle forces (Type I at 100%, 90%, 80%, 70%, 60% and 50% levels). The occlusal angle was varied from 75 degrees to 105 degrees in 5 degree increments for unilateral occlusion at M1, M2 and M3.

Figures 40, 41, and 42 show the effect of occlusal angle (Figure 16) on occlusal force and joint reactions for unilateral occlusion at M1. Similar trends were found for unilateral occlusion at M2 and M3. Occlusal loads are affected very little by occlusal angle, as shown in Figure 40. A minimum is reached at an occlusal angle of 90 degrees. The effect of occlusal angle on ipsilateral and contralateral TMJ loads is quite marked. As the angle is varied from 105 to 75 degrees (unworn dentition to worn dentition), ipsilateral TMJ loads increase, while contralateral TMJ loads decrease.

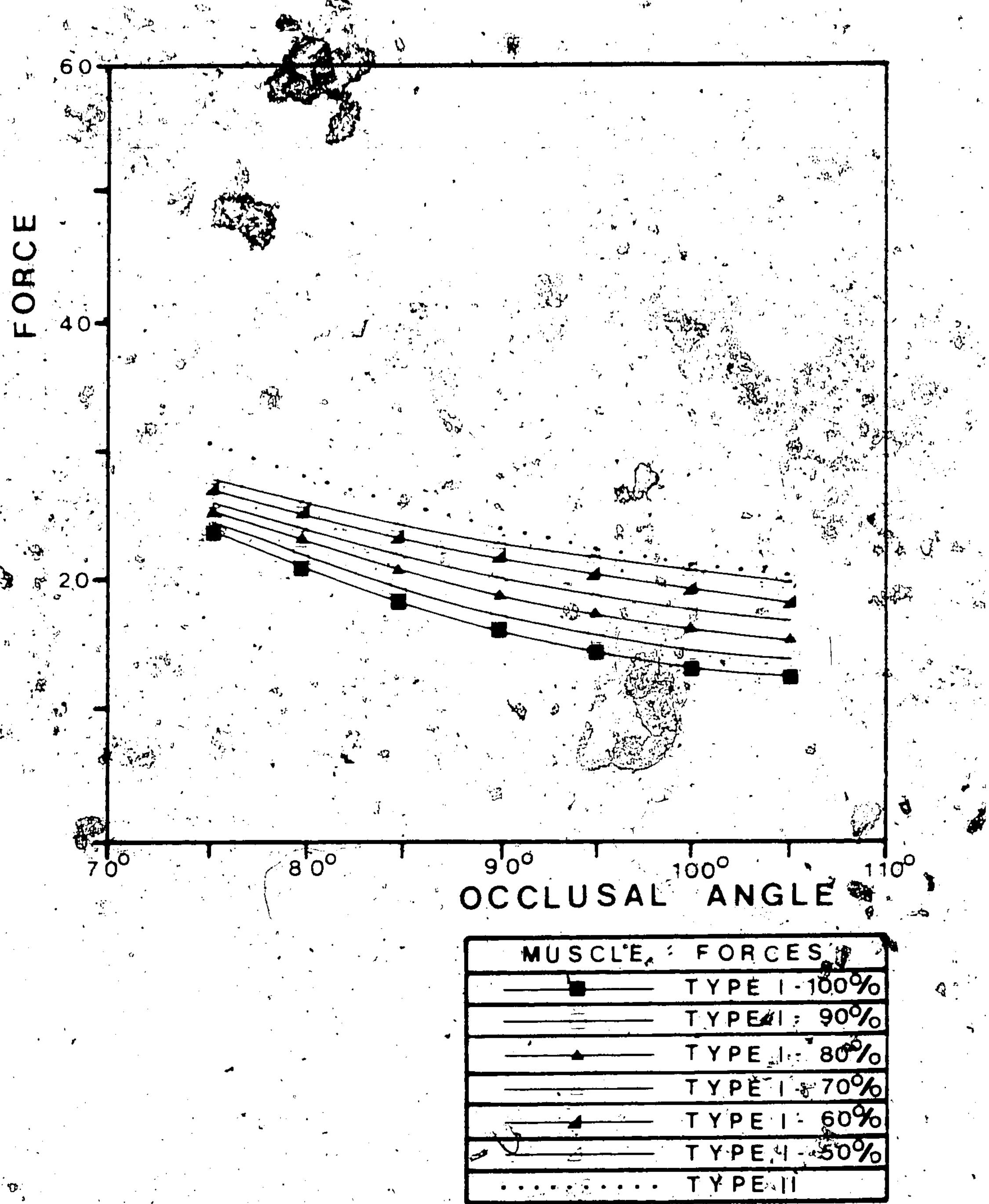
The effect of occlusal angle on TMJ and occlusal loads was determined without adjusting muscle forces with changing occlusal angle. Studies have not been conducted to isolate the effect of tooth wear upon EMG activity of the muscles of



UNILATERAL OCCLUSION AT M1

EFFECT OF OCCLUSAL ANGLE ON OCCLUSAL LOADS

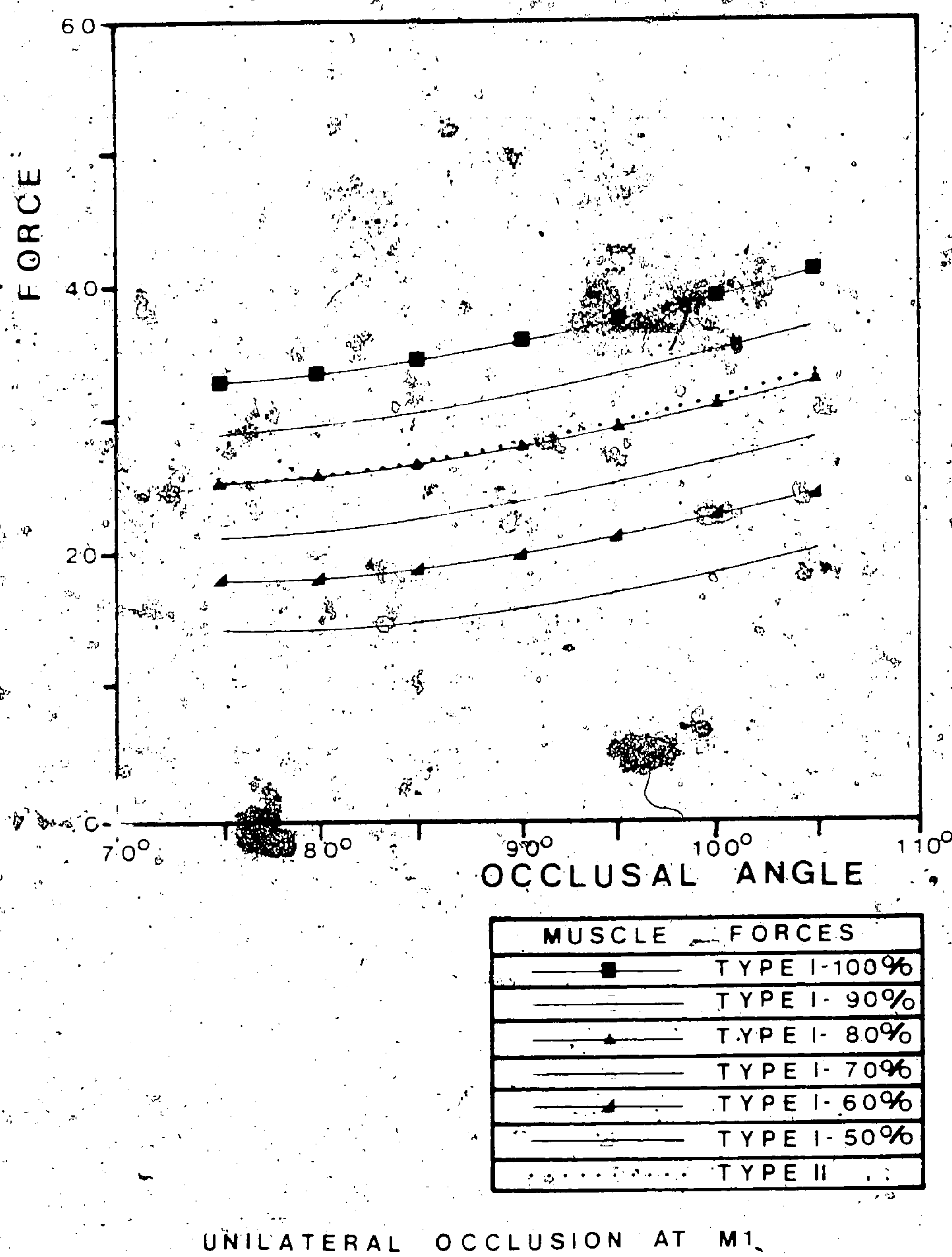
FIGURE 40



UNILATERAL OCCLUSION AT M1

EFFECT OF OCCLUSAL ANGLE ON IPSILATERAL TMJ LOADS

FIGURE 41



EFFECT OF OCCLUSAL ANGLE ON CONTRALATERAL TMJ LOADS

FIGURE 42

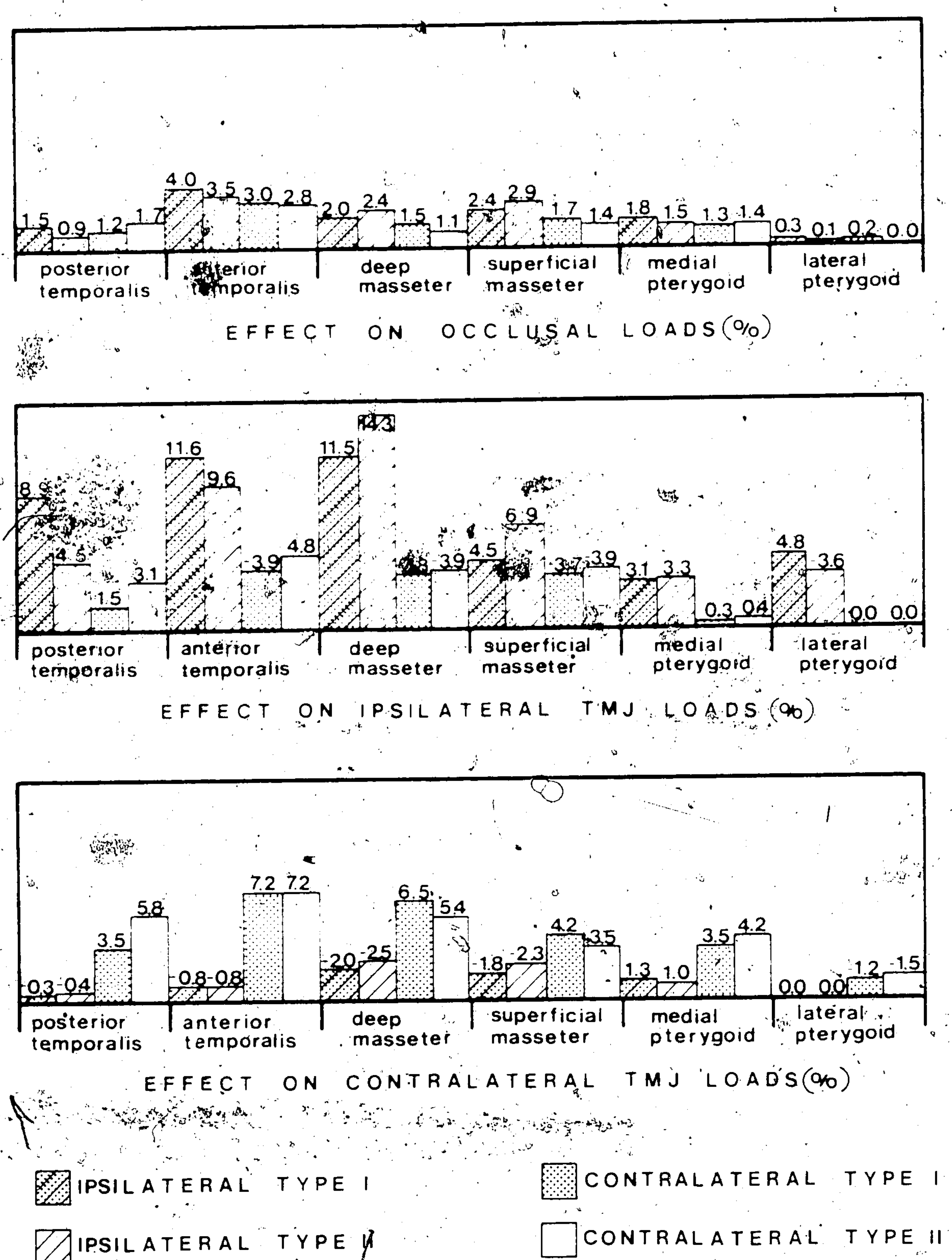
mastication, so it is not possible to determine the validity of not changing muscle forces as the occlusal angle is changed. All that can be concluded is that the mathematical model shows an improved division of the loads on the TMJs with increased tooth wear.

3.7 SENSITIVITY

The mathematical model was utilized to assess the sensitivity of three factors upon occlusal loads and joint forces: (1) the effect of muscle force; (2) the effect of muscle position; and (3) the effect of muscle attachment angle.

Both Type I and Type II muscle forces were tested. All the results from each muscle force scheme were averaged to give an overall view of the sensitivities. The Type I results represent the average of tests performed at M1, M2, and M3 with occlusal angles (Figure 16) varying from 75 degrees to 105 degrees in 5 degree increments and force levels varying from 100% to 50% in 10% increments. The Type II results are the averages of tests run at M1, M2, and M3 with occlusal angles varying from 75 degrees to 105 degrees in 5 degree increments.

The effect of muscle force is shown in figure 43. In this case each muscle force was increased by 20% sequentially and the resulting change in loads expressed as a percentage for both Type I and Type II muscle forces.



MUSCLE FORCE SENSITIVITY

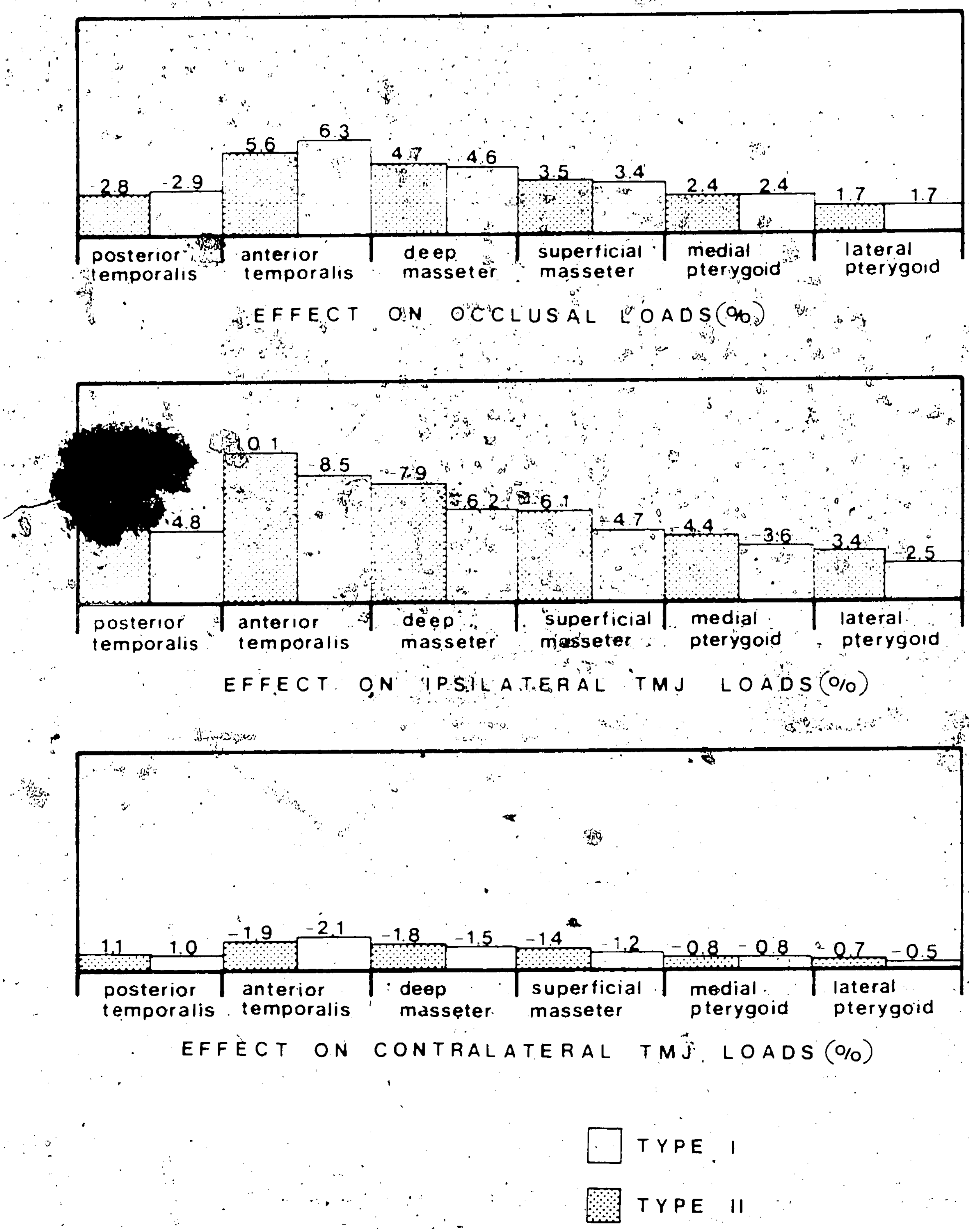
FIGURE 43

The ipsilateral TMJ loads are the most sensitive to changes in muscle force. As one would expect, the ipsilateral TMJ loads are most affected by changes in the ipsilateral muscle forces, while the contralateral TMJ loads are most affected by changes in the contralateral muscle forces. Also, in most cases, changes in the anterior temporalis and deep masseter forces have the greatest impact on the models.

The effect of muscle position was determined by moving the centroid of the area of insertion of an individual muscle by 6.4 mm (0.25 in) in the plane of insertion perpendicular to the muscle force vector. The corresponding change in loads was expressed as a percentage, and these are shown in figure 44.

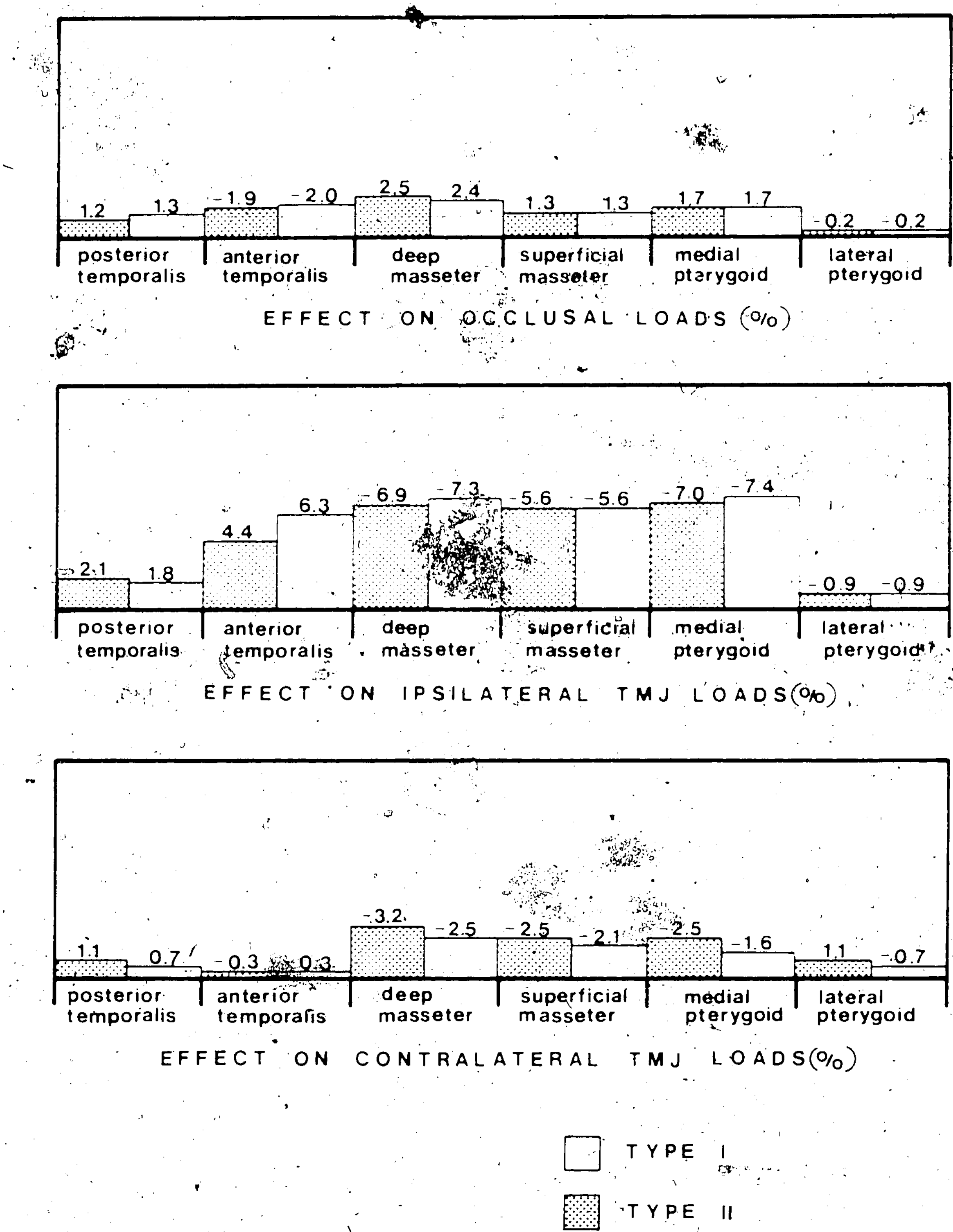
Changing the muscle positions showed ipsilateral TMJ loads to be again more affected than either contralateral TMJ loads or occlusal loads. Occlusal loads were somewhat affected, while contralateral TMJ loads seemed very insensitive. The anterior temporalis and deep masseter muscles caused the largest changes in loads when their positions were varied.

Finally, the effect of muscle angle was examined. The angles at which the muscles inserted on the mandible were changed by 5 degrees in the sagittal plane and 5 degrees in the axial plane for each muscle individually. The change in loads resulting from these changes in muscle angles were expressed as a percentage. These are shown in figure 45.



MUSCLE POSITION SENSITIVITY

FIGURE 44



MUSCLE ANGLE SENSITIVITY

FIGURE 45

As in both force and position sensitivities, the ipsilateral TMJ loads were the most sensitive. The contralateral TMJ loads and occlusal loads were affected to a lesser extent. The anterior temporalis, deep and superficial masseter, and the medial pterygoid muscles all produce fairly significant changes in the resulting loads.

The three sensitivity tests above used the expected variations in the *in vitro* model parameters to determine sensitivities from the mathematical model. The maximum difference between the *in vitro* and mathematical models (37%) was found in ipsilateral TMJ loads. From Figure 43 it can be seen that if all the muscle forces were in error by 20%, the error in ipsilateral TMJ loads could reach ±19.5%. Figure 45 indicates that if all the positions of the muscles were in error, the associated accumulated difference in the ipsilateral TMJ loads would be as large as 33.4%. Should the muscle attachment angle be incorrect, deviations in the ipsilateral TMJ loads may be as great as 28.0%. Thus it is not surprising to see a difference of 37% in the ipsilateral TMJ loads between the models; this difference is certainly explainable with reference to the sensitivities discussed.

4. CONCLUSIONS

4.1 MODEL DEVELOPMENT

In vitro and mathematical models have been developed. The models were found to be in agreement with each other within experimental uncertainties. The mathematical model was further utilized to examine the effect of occlusal angle, occlusal position, and muscle forces on occlusal and TMJ loads. The sensitivities of the calculated loads to the possible variations of the input variables were examined.

4.2 MAJOR TRENDS

4.2.1 TMJ LOADS

The models showed that, in all cases tested, the TMJ was loaded and the contralateral TMJ load was usually larger than the ipsilateral TMJ load for unilateral occlusion. The significance of this is that if the molar teeth on one side of the mandible were lost, forcing the majority of mastication to occur on the opposite side, one TMJ would then be subjected to consistently higher loads than the other. Thus it is important to maintain occlusal contacts in the molar region on both left and right sides to maintain a more even distribution of TMJ loads.

4.2.2 OCCLUSAL POSITION

Unilateral occlusal loads increased as the occlusal position was moved from M1 to M3, while both contralateral and ipsilateral TMJ loads decreased. This result was obtained without adusting muscle forces for different occlusal positions. Some researchers have found that occlusal loads decrease from M1 to M2, while others have found the opposite. This may indicate that in some individuals the muscles of mastication are more developed to apply force for occlusion at M1 than M2 or M3, perhaps as a result of M1 errupting prior to M2 and M3. If muscle forces were to be adjusted to decrease occlusal loads [illegible] M3, TMJ loads would decrease even more sharply as the occlusal position was moved distally. Thus, in order to keep TMJ loads at a minimum, it is important to maintain posterior occlusal contacts.

4.2.3 OCCLUSAL ANGLE

Contralateral TMJ loads decrease as occlusal angle is changed from an unworn dentition to a worn dentition. This result was determined without adjusting muscle forces with changing occlusal angle. Occlusal loads were found to remain fairly constant, while the ipsilateral TMJ loads increased slightly. The net result is that the highest TMJ loads (the contralateral TMJ loads) are decreased without loss of occlusal force as teeth wear, provided the muscle forces do not change with tooth wear. Thus there is a more favourable distribution of TMJ loads with increased wear of the molar teeth.

4.2.4 LATERAL PTERYGOID

The lateral pterygoid muscle (superior head) has no direct effect upon occlusal loads. Its influence on TMJ loads was also minimal in comparison to the other muscles considered. Its major function during unilateral occlusion would appear to be in lending stability to the condyles.

4.2.5 MUSCLE FORCES

Type I and II muscle force schemes both revealed the same trends in occlusal and TMJ loads for occlusal position and angle. However, Type II muscle forces resulted in lower contralateral TMJ loads than Type I, while occlusal loads remained virtually the same. Thus Type II muscle forces were better than Type I muscle forces, as an improved situation is created (ie. the contralateral TMJ load, which was the highest TMJ load, was reduced at the expense of an increased ipsilateral TMJ load). This illustrates the importance of using electromyographical data in evaluating muscle forces.

4.3 AREAS OF FURTHER STUDY

Having developed a mathematical model, further research can now be conducted to determine the effect of antero-posterior inclination of the occlusal plane and dento-skeletal abnormalities upon TMJ loads. The muscle forces used in the model should also be refined by more accurate measurements of muscle cross-sectional area, position and electromyographic activity. The specific area of the TMJ which is loaded during occlusion could also be investigated.

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APPENDIX A

CALIBRATION

1 BITE FORCE TRANSDUCER (BFT)

The BFT was calibrated by applying a known force to the brass plates on the transducer (refer to figure 17), and measuring the corresponding output from both strain gauges. The force was applied in 5 lbf (22 N) increments from 0 to 100 lbf (0 to 445 N) by an Instron test machine[1], and the strain output was measured with a Vishay/Ellis digital strain indicator[2]. Several tests were run to ensure repeatability. The transducer was found to be linear in response, but quite sensitive to alignment (with respect to the applied load direction). Misalignments of approxiamately 5 degrees produced errors as large as 19%. For the force levels measured in the *in vitro* testing, the error in readings taken from the BFT was less than 10%.

2 MUSCLE FORCE TRANSDUCER (MFT)

The MFT was calibrated using dead weights suspended from the transducer through a kevlar sling and weight hanger. Weights were applied in 5 lbf (22 N) increments, up to 30 lbf (133 N), and the corresponding strain output was measured with the Vishay/Ellis digital strain indicator. The

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[1] 50,000 lbf Instron Open Loop Test Machine set in compression mode: 0-100 lbf range.
[2] Vishay/Ellis 20 combined with Vishay/Ellis 21 and Vishay/Ellis 25.

output of these transducers was not linear, and each one required a fourth order least squares curve to accurately represent the response. The repeatability of these transducers was not good, due to a surprising sensitivity to torsional loads. Repeatability was greatly improved if care was taken to eliminate any torsional loads on the transducer. Even so, the MFT's were only considered accurate within 20%.

3 CONDYLAR FORCE TRANSDUCER (CFT)

The condylar force transducers (CFT's) were also calibrated with dead weights. Weights were applied in 2.0 lbf (8.9 N) increments, up to 20 lbf (89N), and each gauge was monitored with the Vishay/Ellis digital strain indicator. Loads were applied at an angle to the transducer in 5 degree increments, with angles varying +/- 30 degrees in the sagittal plane, and +/- 15 degrees in the axial plane. Output of these transducers was both linear and highly repeatable. The maximum error was in the order of 0.3%.