MODELLING OF THE MUSCULAR RESPONSE OF THE HUMAN CERVICAL SPINE SUBJECTED TO ACCELERATION

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ABSTRACT

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A sagittal plane mathematical model for the cervical spine (including T6-T1, C7-C1 and skull) has been developed. In this model the moments due to the weight of the head and neck and the effect of external forces are balanced by forces generated internally by the muscle, the ligament, and the intervertebral joint. With this formulation, the problem is to find a method for distributing the moment between the muscle and the ligament.

Each of the possible solutions has been graded against a mathematical objective function containing the stress experienced by each joint and subjected to the equality constraint (i.e. moment must be balanced) and the inequality constraints (the ability of the muscles and liganents to produce tensile forces only). Using this formulation of the problem, a unique solution that produces a minimum of stress at the intervertebral joints is obtained.

The model is tuned with the use of human experimentation in which volunteers are asked to exert a voluntary pull with their head against a resistance. Electromyographic measurements of various superficial neck muscles have been matched with predicted patterns.

This model has been used to simulate the necks response to high acceleration loading in order to determine the maximum acceleration that the cervical spine can support for different postures. It has been found that the maximum supportable acceleration (i.e. acceleration that would result in any cervical component reaching 2/3 of its limit) depends upon the neck posture and orientation vis-a-vis the acceleration vector. In all cases the system required that the resultant of all forces acting through the occipital-atlas-axis joints be purely compressive.

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LIST OF SYMBOLS

A_{ijk}	-force component i (compression, shear, and moment) at joint-j (C2-C3, C3-C4,...,C7-T1) due to a unit stress in muscle group k (multifidus,scalene,etc.)
K_k	-stress in muscle group k
E_{ij}	-force component i at joint j due to an external load
L_{ij}	-component i at joint j due to the reaction of the ligaments at joint j
J_{ij}	-force component i of the reaction at joint j due to the disc and the facet
$\underline{S}_j$	-unit vector in shear direction of disc level j
$\underline{C}_j$	-unit vector in compression direction of disc level j
$\underline{M}_k$	-unit vector in direction of force of muscle strand k
M_k	-cross-sectional area of muscle
F	-total penalty (objective) function
F_i	-penalty resulting from skeletal structure i (muscle, joint shear, joint compression, ligament)
P_1	-weighting factor associated with the muscles
P_2	-weighting factor associated with the joint shear
P_3	-weighting factor associated with the joint comp.
P_1	-weighting factor associated with the passive resistance of the joint (ie. ligaments)
Nm	-total number of muscles

CHAPTER 1 - INTRODUCTION

1.1 GENERAL INTRODUCTION

Biomechanical modelling is becoming a subject of increasing interest as evidenced by the number of models which can be found in the existing literature. Along with this popularity comes more controversy in view of the number of assumptions that need to be made to model complex biological systems and mechanisms.

The importance of the spine makes it a primary target for engineers. A reliable model would be of great use to designers of equipment who require understanding of the nature of the response of the spine. Some of the classical problems that motivated such modelling effort are 1)pilot ejection , 2)whiplash due to automobile accidents, 3)athletic injuries, 4)the effect of clinical instability, vertebral fusion, description of scoliosis, 5)evaluation of the efficiency of various surgical and non-surgical corrective techniques of the spine, and 6) a method for evaluating and matching a spine to a specific type of work.

Of particular interest is the response of the head and neck to the high acceleration load resulting from pilot ejection. Interest in the problems resulting from pilot ejection has given the main thrust to spinal modelling. This study is restricted to acceleration loads similar to those experience in pilot ejection.

The objective of this study is to develop a system of modelling which is capable of simulating the mechanism of the musculo-skeletal structures of the neck when subjected to forces resulting from sagittal plane loading conditions (i.e. loads acting in the plane of symmetry of the spine).

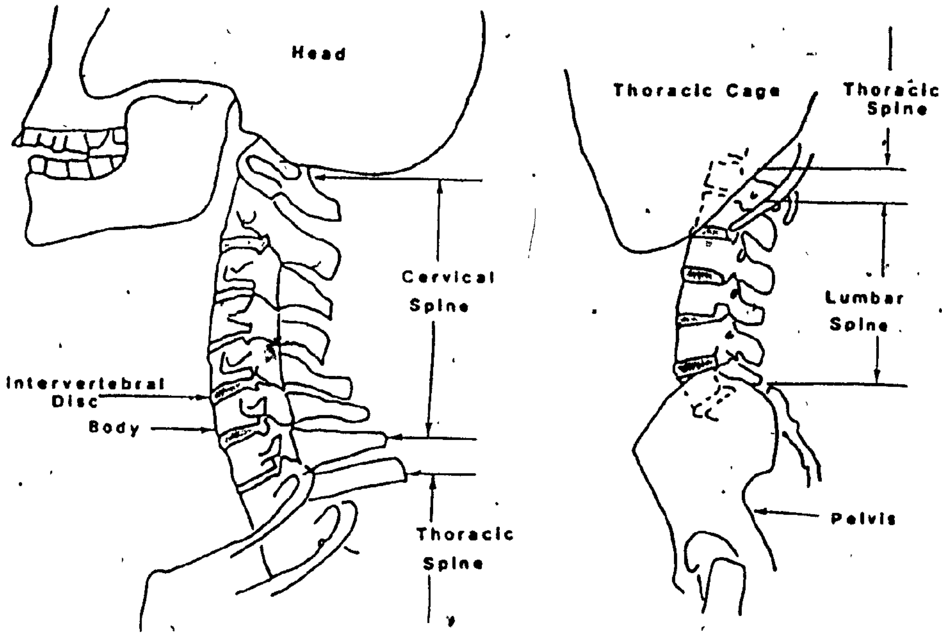
1.2- REVIEW OF THE BIOMECHANICS OF RELEVANT ANATOMY

The vertebral column represents the primary structural member of man. Its mechanical nature can be viewed as a series of segmental bony elements, each poised on a cartilaginous structure which allows adjoining segments to simultaneously possess many degrees of freedom and rigidity as required by the loading conditions.

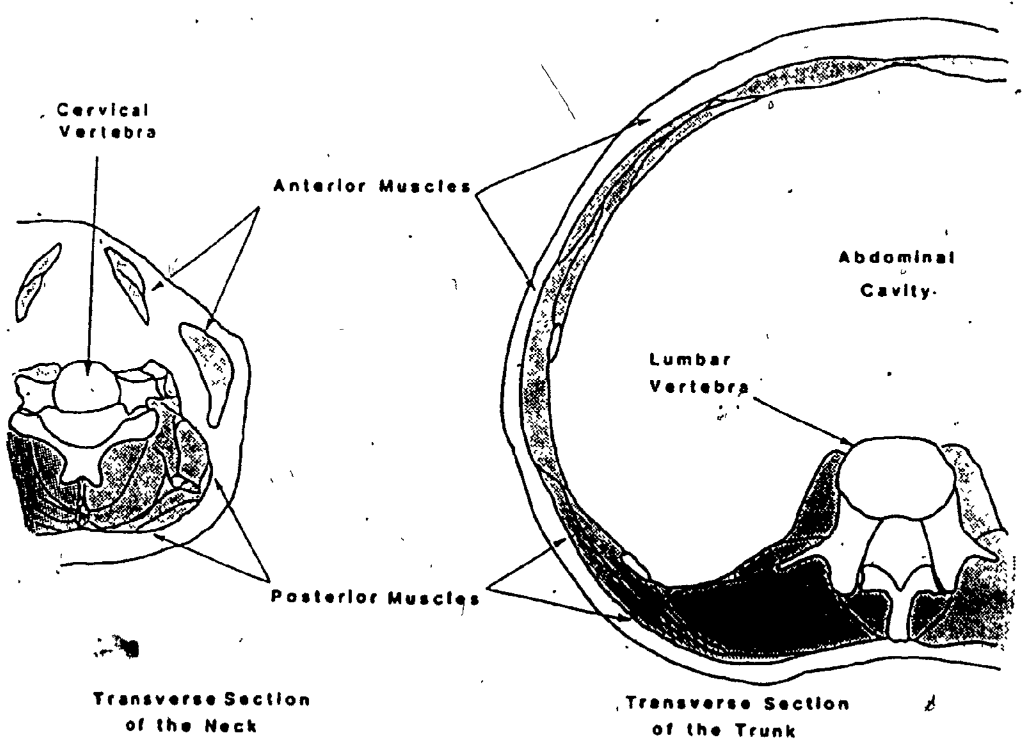
The 24 bones comprising the spinal column are called vertebrae. They are divided into three groups (Fig.1-1a). The first 7 are cervical, the next 12 which bear the ribs are the thoracic, and the remaining 5 are the lumbar. Below the vertebrae, seven bones are united into two structures, the first five form the sacrum and the remaining two the coccyx.

A typical vertebra consists of an anterior segment and a posterior segment. The anterior segment consists of a body that is largely composed of spongy bone surrounded by a thin wall of cortical tissue and capped superiorly and inferiorly by the cartilaginous end-plates of the intervertebral discs.

The posterior segment is made up of the vertebral arch and its accessory processes. Each vertebra body bears paired extensions called pedicles which in turn support the laminae. The laminae and the pedicles form the vertebral arch. Each vertebra carries two pairs of posterior articular processes. They are identified as superior and inferior articular processes. These paired processes bear smooth facets for articulation with the vertebra above and below. The spinous and trans-



a) Lateral view of the cervical spine and the lumbar spine



b) Transverse section through the neck and trunk

Figure 1-1 Schematic illustrating the anatomy of the cervical and lumbar spine verse processes which protrude away from the vertebra body give some added leverage to the muscles which attach to them.

From a biomechanical point of view, the anterior segments will take the bulk of the compressive load while the posterior portion will take the tension by means of the ligaments and muscles which attach to the various processes of the neural arch (Fig. 1-2). With this distribution of tension and compression it is easy to see how the spine acts to support a compressive load and a moment. The role of the articular processes in supporting a compressive load is a subject of debate. For example Nachemson [31] reports that they support as much as 20% of the total compression.

Fresh human cadaver cervical spine specimen are rare and reports generally are based on small numbers. As a result, very little qualitative data on the intervertebral cervical joint is available. The nuchal ligament has been studied and the viscoelastic response to deformation has been demonstrated [15]. There are few hard facts about ligamentous behavior of facet capsular ligaments, the interspinous ligament, the ligament nuchae, the ligament flavum membrane, the anterior and posterior longitudinal ligaments of the neck.

Extensive experimental research has been performed on the lumbar spine and its associated tissues [23, 12, 43, 16, 4, 7, 37, 9, 30, 26, 11]. The properties of the structures in the lumbar region can give insight into the corresponding structures in the cervical region, but the special arrangement of the cervical spine limits the extent to which results obtained from the lumbar spine can be applied to the cervical spine. A comparison of the two regions is illustrated in Fig.1-1.

Motions of the cervical spine joints have been reported in the lit-

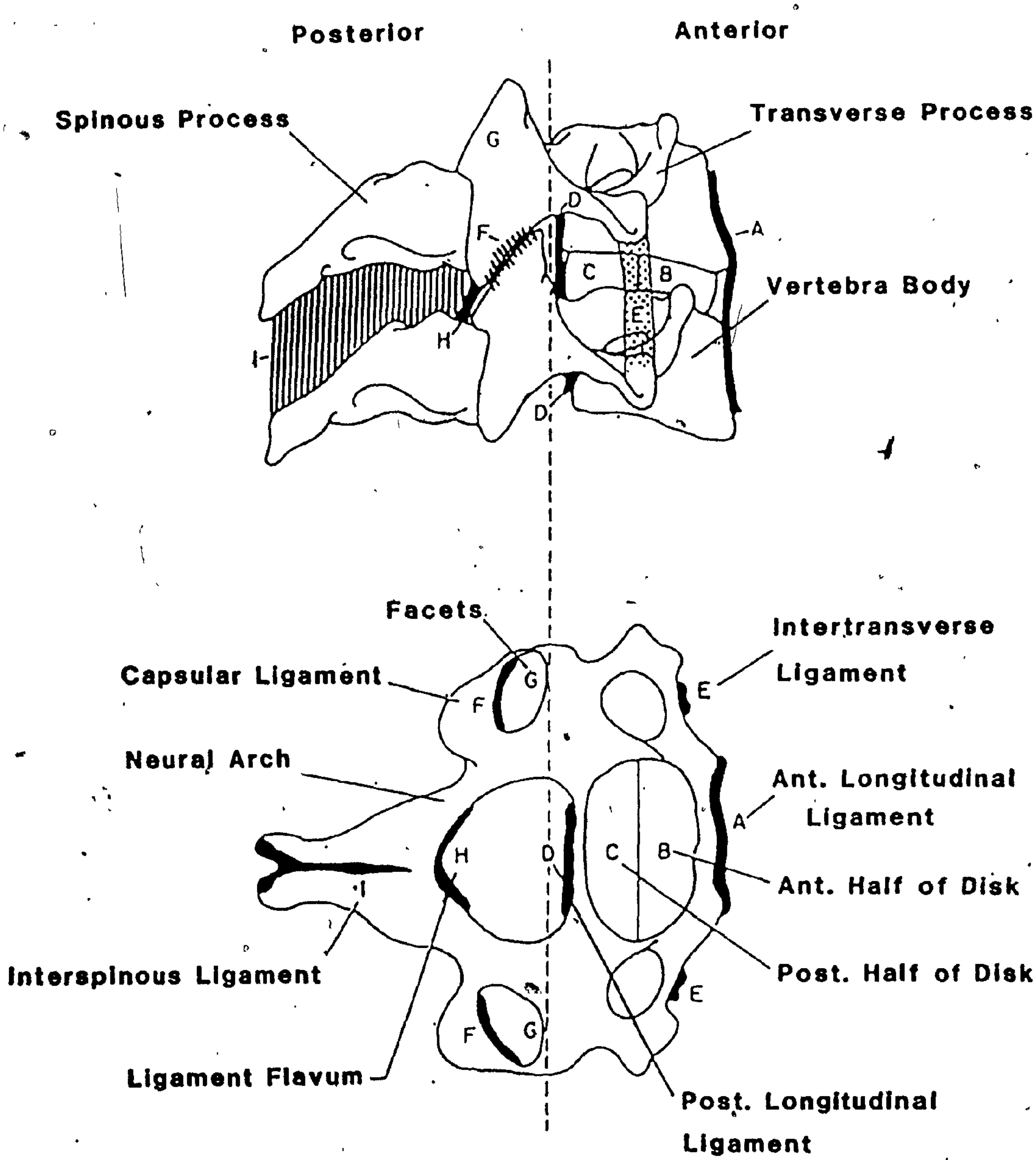


Figure 1-2 Schematic illustrating lower cervical joint (modified from A.A. White and M.M. Panjabi [45])

erature [15,13,1]. The pattern of cervical spine motion has been extremely well demonstrated by Fielding's cineradiography [13] but these studies do not locate the axis of motion or relate the degree of motion to the deforming force.

When an external load is applied to the spine, the fact that the system must be on balance at all times requires that the moment generated by the load be compensated by a moment generated by ligaments and muscles action. In the case of a flexion moment, the maximum moment will be determined by the posterior muscle mass and posterior ligament. For an extension moment the maximum will be determined by the anterior muscle mass and anterior ligament (Fig.1-1b and Fig.1-2).

The electromyographic studies of the muscles associated with the neck are also only sparsely reported in the literature. A good summary of the background of electromyograph (EMG) and the electromyographic studies performed to date is presented by Basmajian [5]. EMG results have been obtained for the longus colli [14,44], the longissimus cervicis [14], and the semispinalis capitis and the splenius capitis [39]. These studies are extremely interesting and give added insight into the function of these muscles. The results of these tests, as with most EMG investigations, are extremely qualitative in nature. As a result, the conclusions which can be drawn from them are very limited.

EMG measurements do not give an absolute measure of the force produced by a muscle. However, the force of maximum muscle contraction has been estimated to be 3.0-8.5 kg/cm^2 [20] and it has been shown that the maximum moment generated by the muscle about the intervertebral joint can be estimated using sectional anatomical specimens to derive muscle areas and radiographs to determine the mechanical advantage of the muscles [10,11]. This procedure has been supported by Rab and Chao [36] who made almost identical estimates.

The mechanical properties of the spinal components are basic to the construction of a model of the spine. Such information provides the constraints within which the model must function. Despite the limited qualitative data available on the biomechanical nature of the cervical spine, rough estimate of the limits of the structures of the neck can be arrived at. These estimates prove extremely valuable in the development and validation of a model of the spine.

1.3 - REVIEW OF MODELLING APPROACHES

The existing modelling approaches have been divided into three categories:

1) the continuum models
2) the discrete parameter models
3) the muscular response models

The first group considers the spine to be a rod or beam. This beam may have varying dimensions but will have the same homogenous material properties throughout. The second group considers the spine as a structure formed by various anatomical elements such as vertebra body, disc, etc., and with different properties being assigned to the different elements. The third group is the modelling approach which is put forward in this study. These approaches are discussed in greater detail below.

1.3.1- The Continuum Models

One example of such modelling philosophy is represented by the work of Latham [27] who in 1958 described the response of the human body to high acceleration in the axial direction (G_z). In this model the spine is represented as a weightless spring with a mass attached at the upper end to represent the body mass and a mass at the lower end to represent the supporting seat structure. With this model it is possible to obtain a dynamic load factor for the spine depending on the assigned spring constant.

The fact that this modelling approach cannot deal with non-axial loads represents a serious limitation since it is obvious from the anatomy that the curvature (lordosis) of the spine will ensure a that purely axial load will never exist. A second limitation is that the model is incapable of singling out the vertebra or disc level which is most likely to fail. This is one of the shortcomings which plague all the continuum models. In order to overcome the problem of not being able to treat non-axial loads, Hess and Lombard [19] introduced the idea of treating the head and trunk as an elastic rod. The base of the rod is considered as fixed and the calculated displacement response of the head during impact accelerations was curve fitted with experimental results.

The Hess and Lombard model was slightly improved upon by Terry and Roberts [40] who modelled the spine as a strictly elastic medium such as a Maxwell type mechanism. Although this model represents an improvement, it can only describe gross body characteristics, and suffers from the requirement that the body mass is evenly distributed along the rod.

Some of the unresolved problems of the above approaches were partly remedied by Liu and Murray [28] in which a Kelvin-Voigt medium rather than the Maxwell medium is used. As with the Hess and Lombard model they represented the spine as an elastic rod and introduced the effects of the head and trunk by capping the rod with a riding mass at the head end of the rod. With this approach they could obtain estimates of the area of maximum axial stress due to a G_z acceleration applied to the hips.

Soechting and Paslay [38] attempted to introduce the effects of the muscles. They did it by lumping the response of the muscles into three parameters: 1) muscle stiffness parameter, 2)neural feedback parameter and 3) a response time delay.

It is apparent that a large number of improvements could be introduced to the model such as an improved representation of mass distribution, damping and varying properties of the column along its length. However, the spine is not a continuous rod and therefore such a model is bound to be inadequate.

1.3.2- The Discrete Parameter Models

The main problem associated with the continuum model is its inability to locate the vertebra level which is most likely to undergo injury when the spine is subjected to extreme loading. The Liu and Murray model claims to locate the region of maximum stress but is unable to specify the precise vertebral level at which it occurs.

This shortcoming is somewhat alleviated by the discrete parameter models. In this modelling approach each vertebra and intervertebral joint is modelled individually, with the vertebra being considered as perfectly rigid and the disc as being deformable. The models differ in how they choose to represent the load-deformation relationship of the intervertebral joint.

One of the first discrete parameter models was presented by Toth [41]. In this model the body mass associated with vertebrae T12 - L5 and the mass below L5 were modelled with eight masses and the intervertebral joints from the sacrum to T11 were modelled with springs and dashpots. Using assumed values for maximum stress, the failure thresholds of individual vertebrae were evaluated. This model was only capable of dealing with axial loads.

A similar approach to that of Toth was taken by Aquino [2] who chose to model the lumbar spine response to G_x accelerations as well as axial accelerations. This was done by representing the vertebrae as rigid bodies and the intervertebral joints with pairs (anterior and posterior) of springs and dashpots. The mass of the head and trunk were considered as a lumped mass.

Orne and Liu [32] represented each vertebra as a rigid body in two dimensional space with three degrees of freedom per vertebra (ie. two degrees for translation in the x and z direction and another one for rotation about the y-axis). The intervertebral disc was considered as a deformable continuum, modelled by a three parameter force deflection relationship. In this manner the model could include the resistance of the intervertebral disc to shear and bending. Different material properties can be assigned to the disc at different levels and the resulting effects on the stress distribution in the spinal column can be evaluated. It is worth noting that this model assumed that the sole supporting structure is the anterior portion of the vertebrae, and that the ability of the spine to resist bending and shear was being assigned to the discs alone. This assumption however does not account for the fact that ligament and muscle tension can significantly affect the bending of the joint.

McKenzie and Williams [29] followed very closely Orne and Liu [32]. They modelled the effects of whiplash by considering only the head and neck (C2-C7). The torso was assumed to be a rigid structure which is restrained by the seat and seat belt. As in the Orne and Liu model, the vertebrae were idealized as rigid bodies and the discs as short uniform beam segments which were represented by a three parameter elastic solid.

Prasad and King [35] extended such an approach to the entire spine. They included the articular facets as a secondary load path in the spinal column. The interaction of the facets was modelled by two springs, one limiting rotation, and the other limiting the relative sliding of adjacent vertebra .

A more recent paper which offers some improvement is that of Belytscho et al [6]. They proposed two major improvements: 1) The abdominal cavity and viscera as hydrodynamic elements stacked in series between the pelvis and T10 level. The contents move vertically and laterally stretching the abdominal wall and transferring the load to the rib cage. 2) The ligaments were included as spring elements which have stiffness only against axial deformation. The introduction of the ligament is a very significant change in the modelling approach since it relieves the disc from the responsibility of absorbing all bending moments.

Other recent models are those of Huston et al [22] and the three dimensional model put forward by Panjabi [34]. As in the other models the vertebra bodies are considered as rigid and the force-displacement relationships are represented by massless springs.

It is important to note that these models represent the muscles as inert masses. Therefore no matter how complex these spring/dashpot models of visco-elastic mechanical behaviour are, they cannot represent the response of the living under the control of the central nervous system. It may be argued that in some extreme situations such as pilot ejection, the musculature can indeed be treated as an inert mass, because the individual does not have the muscular strength to counteract the high acceleration and also that there is not enough time for the neuro-muscular system to react. Nevertheless, even if there is insufficient time for this response to take place, as may occur with impact, the final outcome may be modified by the neuro-muscular system because it may set the initial conditions in the man's favour before the impact begins.

1.3.3- The Muscular Response Models

The major drawbacks of the previous models are: 1) the inability to include the muscular actions, and 2) the fact that many of the values for the spring constants and dashpot coefficients are assumed values due to the difficulty in obtaining reliable test data in the living.

The use of optimization techniques and equilibrium analysis eliminates both of these problems in modelling the musculo-skeletal structure. With this approach it is assumed that the skeleton consists of rigid bodies articulated by joints and held together by muscles and ligaments. The muscles are represented by single or multiple lines of action stretching between their points of origin and points of insertion on the skeleton (Fig.1-3).

The joints are simply subjected to a reaction force and a moment. The two main directions of the reaction force are compression and shear. Compression is defined as the component of the reaction force perpendicular to the bisector of the disc. Shear is defined as the component of the reaction force in the direction of the line formed by the intersection of the bisector of the disc and the mid-sagittal plane.

The optimization problem becomes one of determining the muscle firing combination which balances the applied load and obtains an optimum distribution of stress between the supporting structures. The optimum

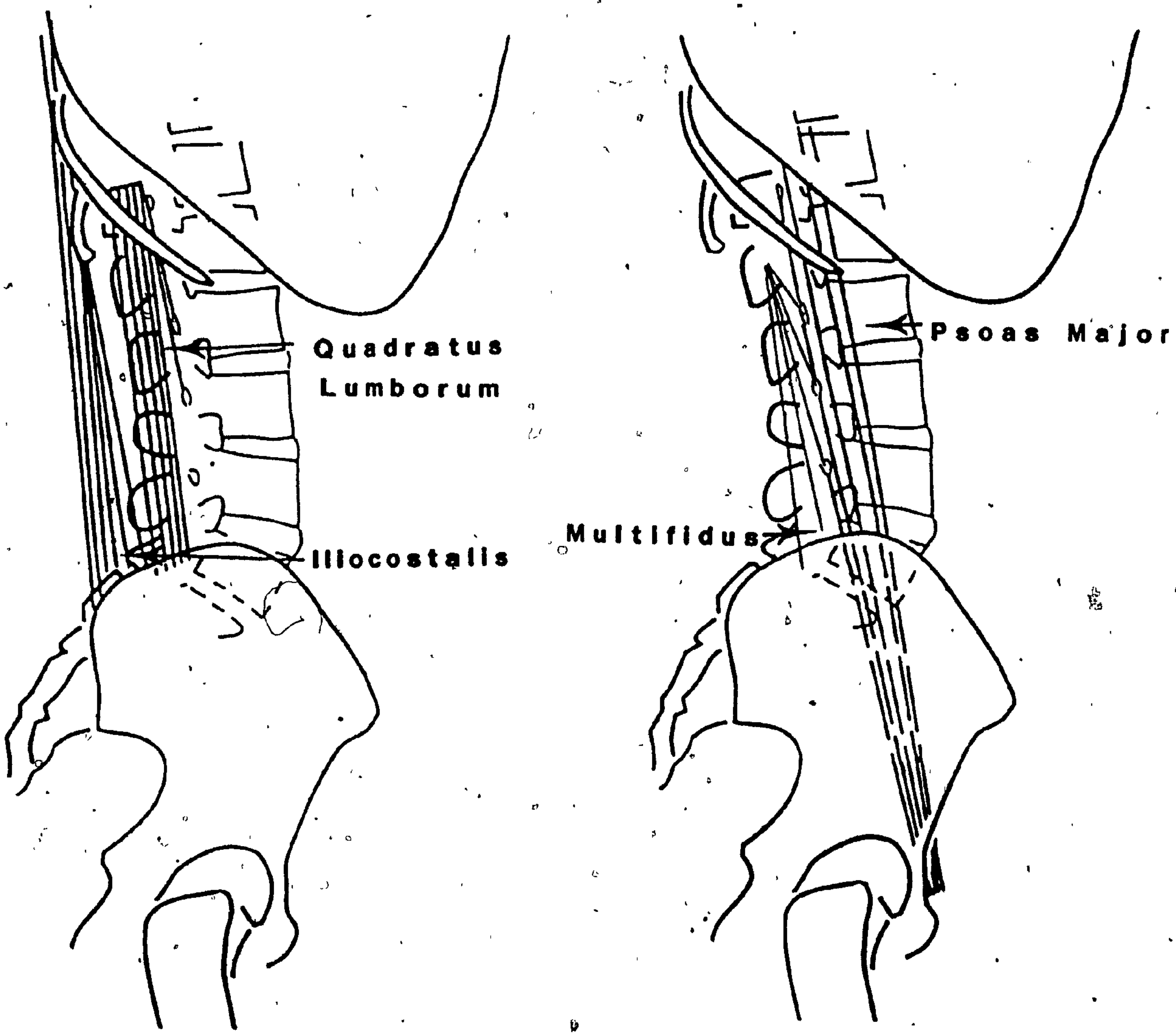


Figure 1-3 Vector representation of some low back muscles (from S. Gracovetsky et al [18])

choice of muscle firing strategy is achieved by choosing an appropriate objective function which is then optimized under the constraints of equilibrium.

This approach essentially recognizes that the ultimate control of the spine is achieved by muscular action, except in extraordinary situations in which the external forces overpower the muscular system. Therefore, these models assign to the muscles the primary responsibility of either balancing the load or to modifying the geometry to allow other structures such as the ligaments to contribute to the equilibrium. By definition, this approach eliminates the need for assumed spring constants and damping coefficients. The major assumption is contained in the definition of what triggers or controls the action of the muscular system. Such assumptions are contained in the objective function to be minimized.

Arvikar and Seireg [3] used the following function:

$$F = \sum K + P_1 \sum M + P_2 \sum R \qquad (1)$$

where $\sum K$ = sum of all muscle forces

$\sum M$ = sum of all reaction moments at the joints

$\sum R$ = sum of all reaction forces on the joints

P_1 and P_2 are weighting factors

This objective function is linearly dependant on the stress levels in the joint and muscle. The joint is treated as a black box which contains the intervertebral disc, articular facets and the ligaments. In this model no attempt is made to determine the distribution of stress between the passive compressive elements of the joint (the disc) and the passive tensile elements of the joint (the ligaments). Also no attempt is made to account for the difference in the joint's ability to support the loads in the two main directions.

The choice of this particular objective function is necessarily arbitrary, and can only be justified by the consequences of the function on the model behavior. Hence this type of model requires extensive simulation to prove that the calculated muscular coordination matches the observed muscular coordination. Such a validation procedure is feasible because the model generates the muscular activity itself, which can then be verified by electromyographic measurements.

With this approach Arvkar and Seireg used this model to simulate a subject in the seated posture with 1) no acceleration, 2) a forward acceleration, and 3) a backward acceleration. He was also modelled in a stooped position of 52 degrees with and without a weight. However no experimental studies were made to validate this model.

The only other model in the literature which incorporates the active response of the muscles is that of Gracovetsky et al [17,18]. This model simulates the stress distribution in the low back of a weight lifter performing a dead lift. The muscle tensions were determined by obtaining the optimum stress distribution while balancing the externally applied load. The model predicted that the optimal lifting sequence is achieved when the stress is equalized at all intervertebral joints. The objective function used is of the following form:

$$F = P_1\sum K^2 + P_2\sum S^2 + P_3\sum C^2 + P_4\sum M^2 \qquad (2)$$

where $\sum K^2$ = sum of the square of the muscle firing densities

$\sum S^2$ = sum of the square of the shear at each joint

$\sum C^2$ = sum of the square of the compression at each joint

$\sum M^2$ = sum of the square of the reaction moments at each joint

P_1, P_2, P_3 and P_4 are weighting factors

This quadratic objective function accounts for the difference in the joints ability to produce a reaction force in the two principal directions. The model also accounts for reaction force produced by the ligaments.

Validation for the model is obtained by showing that during the execution of a dead lift it is sufficient to minimize the stress at all intervertebral joints in order to generate a sequence of muscle activity which has the following properties:

1- It reproduces faithfully all known muscle averaged EMG patterns.

2- The muscle power required for the lift has been confirmed in vivo.

3- The biological limits of the tissues are not exceeded.

4- The maximum voluntary effort (400lbs lift) requires only 2/3 of the available resources.

The advantage of the use of optimization technique is that it yields stress values at joint levels without making gross assumptions about the force displacement relationship of the joints. The apparent arbitrariness of determining a suitable objective function may be seen as a positive feature since its gives additional insight into the nature of the spinal mechanism, as well as the fundamental laws that govern the use of spinal resources.

1.4 OBJECTIVE AND CONTRIBUTION OF PRESENT STUDY

1.4.1 Objective

Examination of previous modelling approaches revealed that the continuum modelling approach was incapable of determining the specific structure which would limit the spine in supporting a load. The discrete modelling approach is forced to rely on assume parameters when insufficient test data is available. A greater problem of the discrete parameter model is its inability to determine the contribution of the muscles. This second limitation is perhaps the most serious since it is felt that the muscle will play a very critical role in the supporting of a load or the execution of a movement.

The objective of this study is to develop a modelling approach which is capable of describing the muscular response of the cervical spine with sufficient accuracy that comparisons may be made between the simulated muscle response and the electromyographic activity obtained experimentally. At the same time it is desirable to limit the complexity with which the passive spinal structures such as the joint and the ligaments are described so as to avoid the necessity of introducing a large number of assumed parameters as was done in the discrete parameter spinal modelling approaches of the past. It is felt that these two features may be incorporated into an optimization model similar to the one of Gracovetsky et al [17,18] by defining the optimization problem in the appropriate fashion.

The main contribution of this study is to apply the principles developed in the work of Gracovetsky et al in modelling the lumbar spine and apply them to the cervical spine. In addition, EMG experimentation is more closely integrated into the modelling procedure.

1.4.2 The Importance of the Objective Function

Wolff's law [45] characterizes the response of the bone when subjected to stress. It states that bone is added where it is needed and removed where it is not. Hence the very shape of the bone, including its internal architecture, is determined by the stress it experiences. Ultimately, Wolff's law is responsible for the shape and size of all the bones and therefore the vertebra.

All the bone of the vertebra must be stressed equally, at least on an average basis and at least to some non zero minimum level. Consider the possibility of one section of a vertebra being subjected to greater stress levels relative to another section. It would then become necessary for this section to grow extra bone. This would result in a lowering of the average stress level experienced at this section of the vertebra. The same argument can be applied to the case of an understressed section of the vertebra.

It can be inferred from Wolff's law that the stress is equalized throughout the intervertebral joint. Because the tissues are identical regardless of the spinal level, it follows that the stress in all spinal skeletal levels must be equalized. The most economical use of material would require that any task be executed in such a way that the stress be equalized and minimized. Therefore a precise mathematical formulation should describe the spinal mechanism using a distributed parameter approach coupled with the requirement that the solution does minimize and equalize the stress through the spine.

The objection to such an approach is that we do not know how to distribute the load and furthermore the mechanical properties of the spinal members are not known in sufficient detail to make such description feasible. What should be done is describe the spinal mechanism using a level of complexity that does not exceed the available experimental data. Therefore the formulation of the problem must be simplified. For example the distributed forces at the vertebral end plates can be replaced by a resultant vector. Similarly the muscles and ligaments were represented as a finite set of vectors with points of origin and insertion.

In the simplified model of the lumbar spine, which required the stress to be minimized, validation is achieved in two ways:

1- By obtaining a solution that is found to equalize the stress. Here the optimization procedure itself is validated since it is not obvious that stress minimization implies stress equalization. In other words Wolff's law is used to validate the procedure, namely the objective function.

2- By obtaining a solution that explained all available experimental data on the lumbar spine. Here the spine experiments are used to validate the mathematical representation of the spine itself, namely the model.

1.4.3 Approach

This study is divided into four sections. The first part of the study consists of reducing the relevant anatomy of the neck to an appropriate numerical form. The portion of the spine which is under study extends from the head down to the first thoracic vertebra. Since some of the muscles which traverse the cervical joints extend down to as low as the sixth thoracic vertebra and the shoulders, these skeletal structures are also be considered.

The second part of the study consists of an electromyographic investigation of the superficial muscles of the neck which are accessible to surface electrodes. This experimental investigation consists of a preliminary set of experiments which is followed by the main portion of the investigation. The purpose of the preliminary investigation is to determine what information can be derived from the EMG signals. These results are then used to design the main investigation.

The third part of the study is the development of the model and the simulation of the experiments performed in the main portion of the EMG investigation. This consists of developing the objective function, the equation which represent the constraints within which the model must function, and the grouping of muscle vector. The final form of the objective function and the muscle groupings are based on the results obtained from the EMG investigation.

The final part of the study is the use of the model developed in the previous section to simulate the cervical spine subjected to acceleration loading conditions similar to those experienced in pilot ejection.

The principles developed by Gracovetsky et al have only been applied to the lumbar spine of a subject performing a dead lift or to tasks which results in a load similar to those experienced in the dead lift. The extension of these principles to the cervical spine and different loading conditions can be seen as further validation of them.

CHAPTER 2 EMG INVESTIGATION OF THE HUMAN CERVICAL SPINE

2.1 OBJECTIVE

The objective of the experimental investigation is to measure the firing pattern of cervical spine muscles of volunteers performing a specific task. The muscular pattern is measured from EMG activity collected by surface electrodes. This muscular pattern is then compared elsewhere with the muscular pattern calculated by the model.

The overall investigation is divided into two parts. The first part, the preliminary investigation, determines the accessibility of the superficial muscles of the neck and the repeatability and consistency of the data in order to evaluate and design the main experimental procedure. To this end, several volunteers are examined using an eight channel EMG recording equipment. The second part, the main investigation, is carried out with substantially more resources, such as on-line computer data acquisition and processing.

2.2 PRELIMINARY INVESTIGATION

The neck muscles are examined to determine which superficial neck muscles could be monitored using miniature surface electrodes. Tests are performed on a number of subjects who are asked to maintain a set posture while increasing load is applied by hand to their head. This portion of the study is only qualitative in nature and did not follow a predetermined pattern.

The procedure followed is to palpate the muscle (if possible) and place a pair of miniature silver-silver chloride electrodes on the region of the skin over the muscle with the use of adhesive collars. A single common reference ground electrode is placed behind the earlobe. A colloidian glue is used to facilitate electrode adhesion in the hairy regions of the neck (ie. semispinalis capitis and splenius capitis placements).

Myoelectric activity is amplified by the eight channel Beckman R-611 electromyograph.

Electrode placement - A number of muscles are monitored for EMG activity in a number of loading configurations. In all cases, EMG recordings are noted to show the level of activity during backward extension against resistance. The results of these tests are summarized below and in Table 2-1.

1) Spinalis and Semispinalis Capitis: The placement of the electrodes for this muscle is 2cm below the occipital bone and 2cm lateral to the midline (8). It is found that activity can be observed during extension against resistance and that this activity would increase with increasing effort.

2) Splenius Capitis: The splenius capitis electrode placement is 3cm below the mastoid process and 3cm lateral to the midline (8). This particular placement ensures that little or no activity is obtained from the semispinalis capitis. It is found that activity can be observed from this muscle during extension against resistance and rotation against resistance.

MOTION AGAINST RESISTANCE

Muscle	Extension	Flexion	Rotation	Lowering Jaw	Respiration
Semispinalis Capitis	activity	no activity	no activity	no activity	no activity
Splenius Capitis	activity	no activity	activity	no activity	no activity
Sterno-mastoid	slight activity	activity	activity	no activity	no activity
Omohyoid	slight activity	activity	no activity	activity	no activity
Sternohyoid	no activity	activity	no activity	———	———
Scalene Anterior and Posterior	no activity	———	———	———	———

Table 2-1 Results from preliminary EMG investigation to determine which neck muscles are active during resisted extension

3) Sternomastoid: The electrode placement for this muscle is 4cm from the mastoid process along the line of the muscle. The placement of this pair of electrodes presented no problems due to the fact that the muscle is very superficial and easy to palpate. Activity is observed during rotation and flexion of the head against resistance. Little or no activity is observed during extension against resistance.

4) Omohyoid: The electrodes are placed slightly above the clavicle and about 1 to 2cm lateral to the clavicular attachment of the sternomastoid. Activity can be observed during opening of the jaw against resistance. No activity is observed during rotation of the head against resistance.

5) Scalene Anterior and Posterior: Two placements are used for this muscle. The first is slightly above the clavicle and slightly anterior to the lateral margin of the trapezius. The second is the same as that used for the omohyoid. Activity can be observed from both placements during breathing. No activity is observed for the posterior placement during extension against resistance. Activity is observed for the anterior placement and is thought to be due to the omohyoid.

6) Sternohyoid: A single pair of electrodes is placed slightly below the hyoid bone. Activity can be observed during flexion against resistance and no activity is recorded during extension against resistance.

Procedure: The existing equipment consists of only 8 EMG channels (8 pairs of electrodes). Since it is desired to monitor the muscles bilaterally, a choice must be made as to which muscle are to be monitored. Based on the results described above, it has been decided to have the subject extend the neck against resistance with electrode placements over the semispinalis capitis, splenius capitis, sternomastoid and omohyoid muscles.

The other muscles have been eliminated on the assumption that they would produce little activity during this isometric exercise. Since this may not be the case, these muscles will be examined later in the integrated study.

The loading procedure consists of a horizontal load applied to the back of the head (producing sagittal plane effects only). The main purpose is to determine, as that load increases:

1) to what degree (if any) the sternomastoid and the omohyoid muscles are being recruited, and

2) the manner in which the splenius capitis and the semispinalis capitis muscles are being recruited under the same conditions.

The applied load (Fig.2-1) can be developed effectively by having the subject extend his head against a resistive force. When this happens the subject is said to increase his extensor moment. The combination of the applied and the gravitational load (ie. the weight of the head and neck) gives rise to the resultant load. It is important to

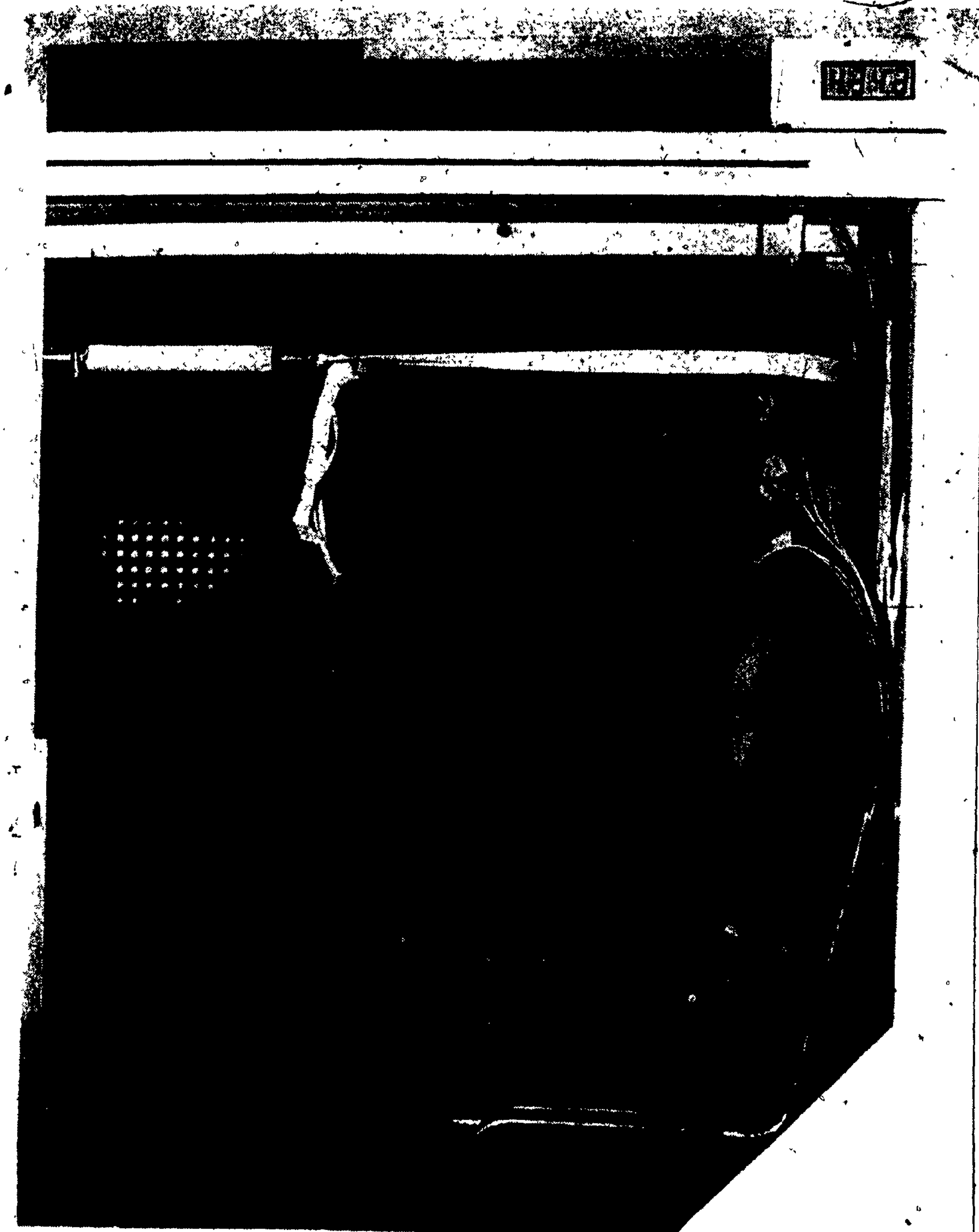


Figure 2-1 Lateral view of experimental loading procedure

note that as the applied load increases, the resultant load changes direction. This change in direction is the feature of the loading sequence which elicits changes in the muscle firing pattern.

Seventeen healthy adults have been examined (10 males and 7 females) ranging in age from eighteen to seventy-nine. Paired miniature silver-silver chloride electrodes are placed bilaterally on the sternomatoid, splenius capitis, and omohyoid muscles. The semispinalis capitis muscle is recorded unilaterally due to an insufficient number of recording channels.

The relative position of the omohyoid with respect to the clavicle varied considerably from one subject to the next. In three cases electrode placement proved to be impossible because of its complete obscuring by the clavicle. Due to the proximity of the sternomastoid muscle, electrode placement for the omohyoid requires verification. A simple opening of the jaw has proven to be sufficient for this purpose.

The loading apparatus consists of a typical strain gauge device mounted horizontally with a comfortable sling attachment to accommodate the head. Seated in a chair, the subject is asked to pull back against the sling, thereby increasing his extensor moment (Fig.2-1). The shoulders are relaxed with the arms hanging loosely at the side. An idealized view of the loading arrangement is depicted in Fig. 2-2.

A sequence of photographs are taken of each exercise at 4 frames per second. Onto these, lines are constructed, a) from the outer canthus of the eye to the external auditory meatus, and b) from the sternum to the prominent vertebra C7. By measuring the change in angle formed by these two lines, it is possible to detect any change in geometry during the performance of the task.

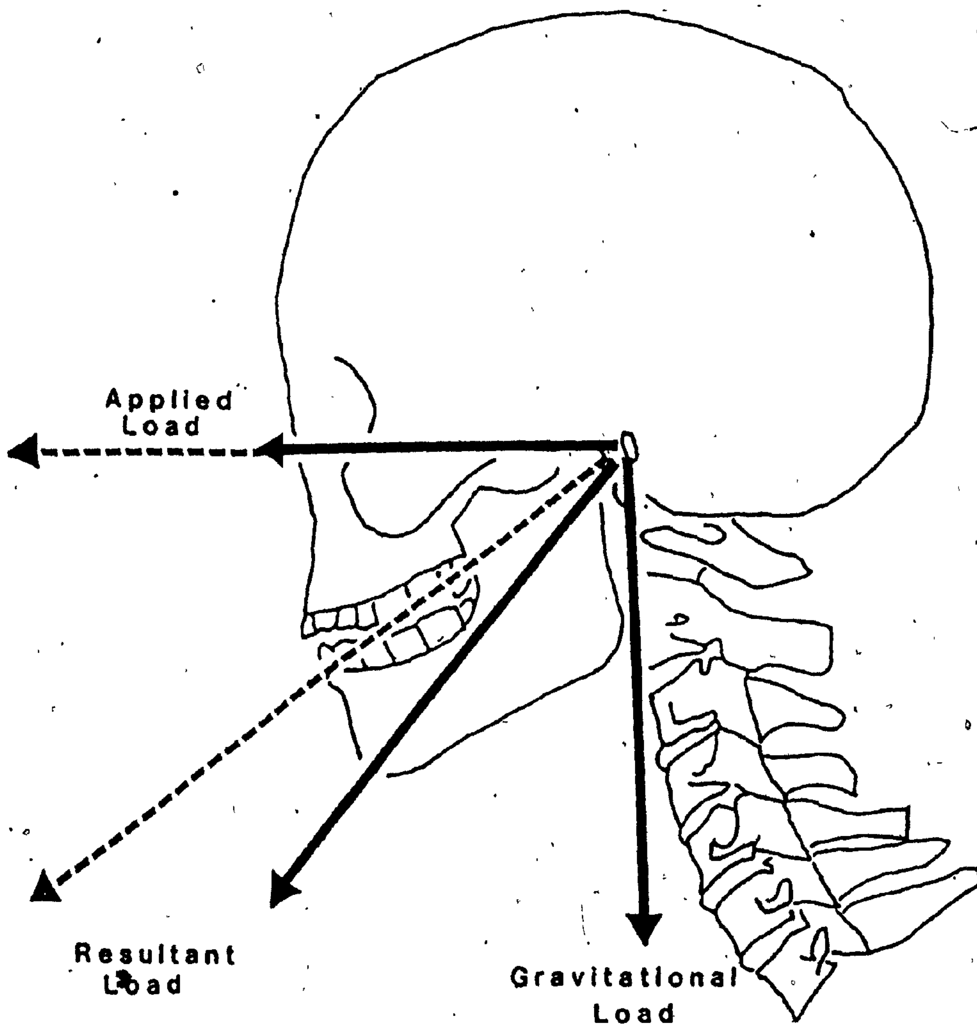


Figure 2-2   Idealized view of experimental loading procedure

RESULTS Integrated EMG signals from the sternomastoid and omohyoid muscles are graded as a percentage of their "assumed maxima", determined by a resisted forward flexion for the former and a resisted jaw opening for the latter.

The histograms (Fig.2-3) show the distribution of the total number of muscles examined with respect to their assumed maxima. Because the sternomastoid and the omohyoid have been recorded bilaterally, they each contains 34 readings of muscle output. The histogram for the omohyoid muscle shows a considerable spread in the results. The histogram for the sternomastoid muscle indicates the activity of this muscle to be confined primarily in the range of 1-10% maximum. Twenty eight out of 34 muscles exhibited activity in this range. Mean values for the omohyoid and sternomastoid muscles are calculated at 25% and 4% of their assumed maxima respectively.

In all 17 subjects the following observations are made concerning the semispinalis capitis and splenius capitis muscles. As the extensor moment is increased (ie. as the subject pulls harder), the semispinalis capitis is recruited immediately. However, the splenius capitis initiates its rise in activity only after a certain length of time (Fig.2-4a and Fig. 2-4b). It is further noted that the splenius capitis muscle is recruited at approximately the same value of extensor moment regardless of the rate at which the subject is pulling.

In all 17 cases, measurements taken from the photographs failed to detect any change in the angle defined above. Hence it has been concluded that there are no significant changes in geometry.

This set of EMG tests is intended as a prototype to allow us to design the main investigation protocol. The results of this preliminary

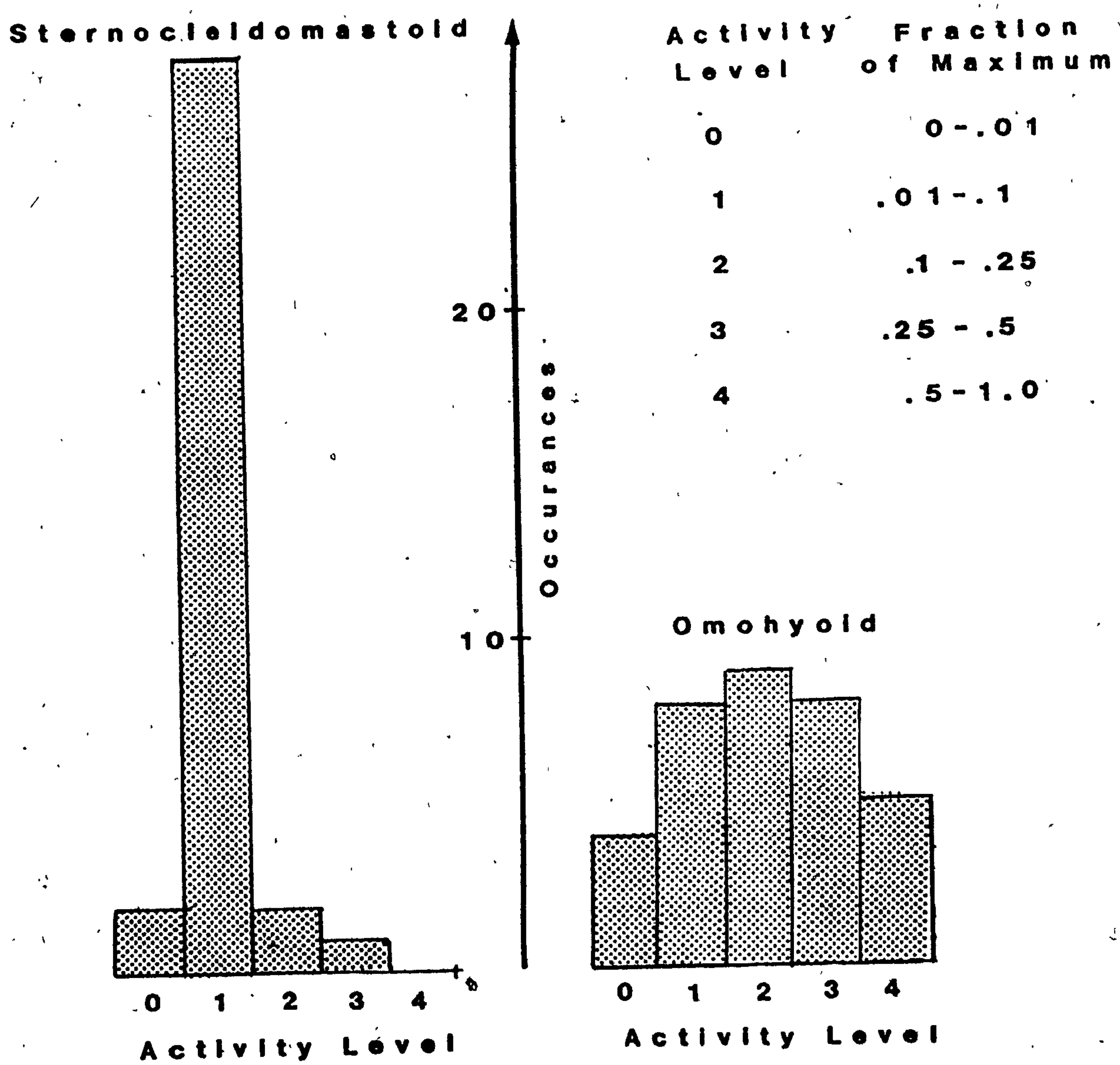
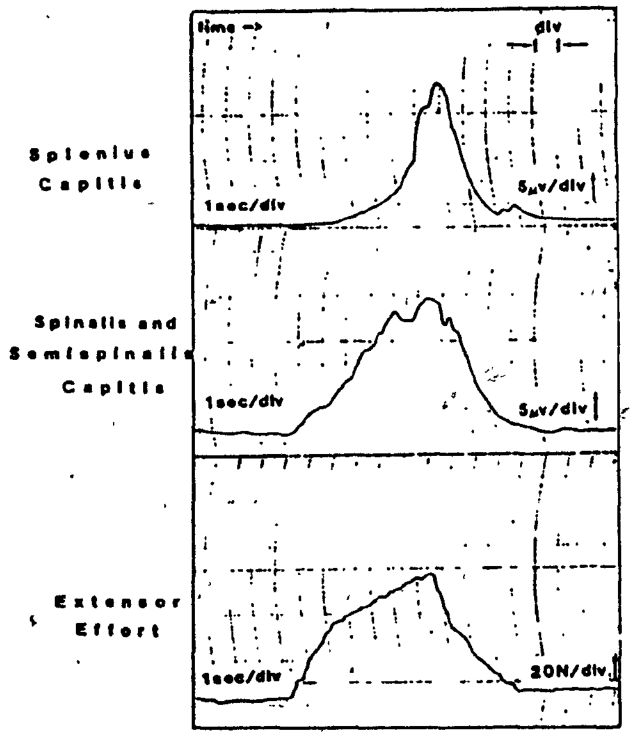
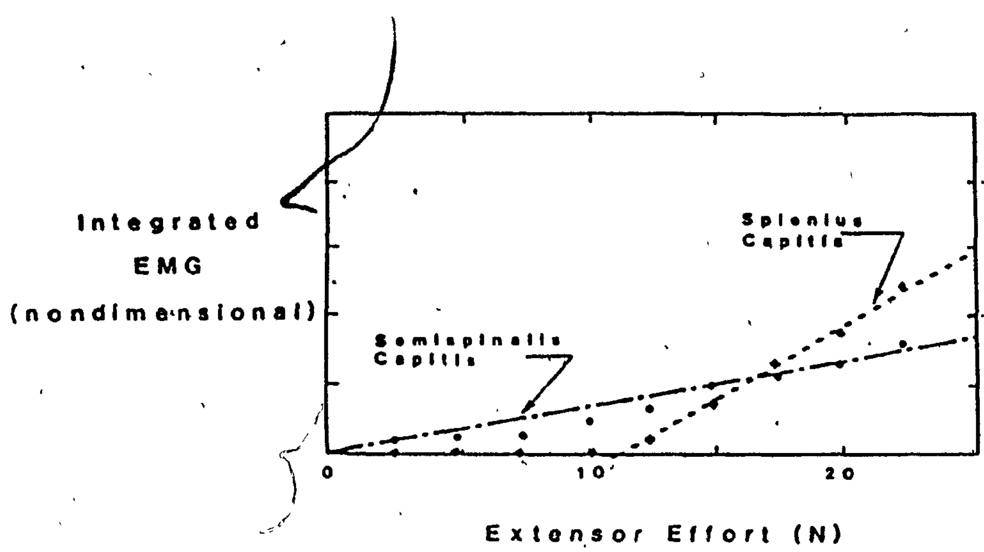


Figure 2-3 Histogram of results for sternomastoid and omohyoid muscles



a) Integrated EMG versus time



b) EMG activity versus extensor effort

Figure 2-4 Experimental muscle firing pattern for semispinalis capitis and splenius capitis muscles investigation suggest a number of modifications which can be made to the apparatus and the test procedure:

1) Since the extensor effort at which the muscle activity is initiated is of primary interest, the load cell must have sufficient sensitivity to allow accurate determination of this value.

2) The apparatus on which the load cell is mounted must permit varying directions of loading.

3) Due to the unexpected results of the sternomastoid muscle, recordings of the EMG activity in the scalene, sternohyoid and the trapezius muscles is necessary.

2.3 MAIN INVESTIGATION

Changes to the apparatus: The load cell and the load cell mounting apparatus are changed as follows:

1) The strain gauge rod is replaced by a universal flat load cell with a full scale of 111 Kg (250 Lb) and a sensitivity of 0.1% of full scale. The load is applied to the load cell through a bolt passing through a threaded hole in the center of the load cell.

2) The resistance to the subject's extensor effort is provided by a strap which passed around the subject's head and attaching to the strain gauge bolt.

3) The load cell is bolted down to a mounting device which rotated on a ball joint to permit rotation in the vertical and horizontal planes. This load measuring device is bolted to the same rigid frame which is used in the preliminary investigation. The height of the load cell can be adjusted to produce different lines or angles through which the applied load can act.

Loading procedure: The electrode placement has been described in the preliminary investigation. The subject is asked to attempt to pull his head backwards against the restraining strap. This produces a flexion load which is monitored with the load cell apparatus. The analog signals resulting from the load cell and the amplified electromyographic signals are 1) passed through an active analog band pass filter which passes a frequency band of 10Hz to 250 Hz and 2) digitized (12 bits) at a rate 1000 samples per second and stored. The analog filtering adequately attenuated the biopotential signal resulting from the heart (ie. the ECG) and eliminates high frequency noise. Since the highest frequency signal to be digitized is 250Hz, a sampling rate is more than adequate to obtain the require information from the signal.

The test performed on each of the volunteers consists of 5 tasks or acquisitions. These tests are as follows:

TASK #1 Electrodes are placed bilaterally over the semispinalis capitis, splenius capitis, omohyoid and the sternomastoid. The height of the load cell is set at a level which resulted in the restraining strap sloping downwards at about 5-10 degrees away from the subject's head. The subject assumes a normal upright neck posture and gradually draws back against the strap until he achieves his maximum extensor effort.

TASK #2 The same electrode placement and load cell height as in task 1 with the subject pulling in the flexed neck posture.

TASK #3 The same electrode placement and load cell height as in task 1 with the subject pulling in the extended neck posture.

TASK #4 The same electrode placement as task 1 with the load cell lowered to produce a 45 degree downward slope away from the subject in the restraining strap. The subject assumes a normal upright neck posture while drawing back on the restraining strap.

TASK #5 The same electrode placement and load cell height as in task 4 with the subject pulling on the strap in the flexed posture.

Electrodes can be placed bilaterally over only four sets of muscles. It is therefore necessary to perform a separate examination of the remaining superficial muscles. The electrodes on the omohyoid and sternomastoid muscles are removed and replaced bilaterally over the posterior scalene muscle and unilaterally over the trapezius and sternohyoid muscles. The procedure of task 1 is then repeated.

Five volunteers are used in this investigation with each of the volunteers being tested on three different occasions to give a total of 15 tests. A small number of volunteers being tested on more than one occasion is deemed appropriate for two reasons:

1) Poor electrode placement or a high level of interfering signals being picked up during the test can make it difficult or impossible to obtain a good estimate of the desired information.

2) It is informative to know if observations made on a subject in a given test can be observed in subsequent tests. In this way it is possible to determine if variations in the level of extensor effort required to recruite the various muscles is a function of the individual being teseted or whether there is a certain amount random variation which are a function of the EMG testing.

Results: The results of the preliminary investigation suggest that the information which is the most useful in tuning the model is the level of the extensor effort required to recruit each of the muscles being examined. These values are estimated by processing the original

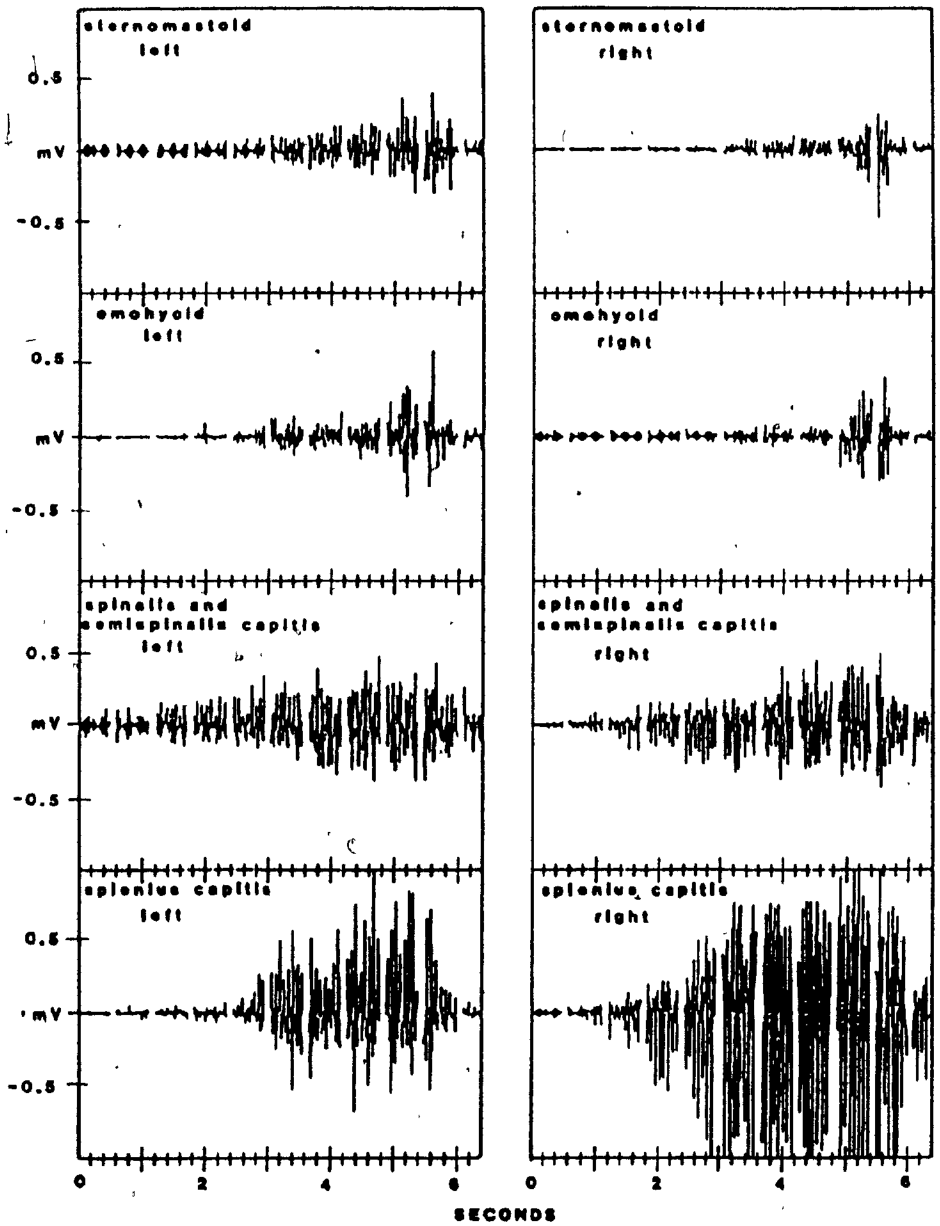


Figure 2-5 Sample EMG data for volunteer #1 performing TASK - 1

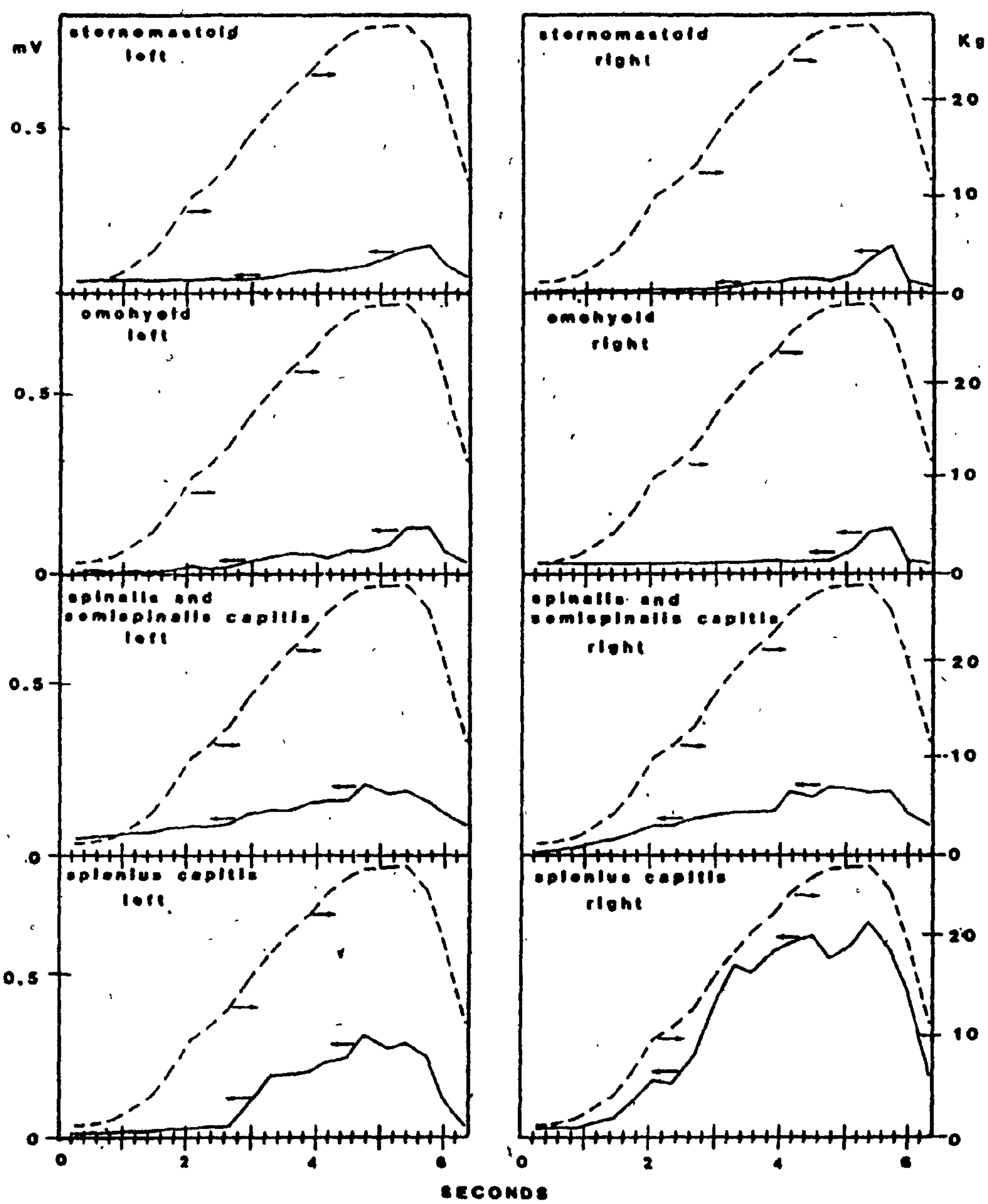


Figure 2-6 Sample IEMG results for volunteer #1 performing TASK - 1

----- load cell output read from the scale on the right

——— IEMG results read from the scale on the left

EMG signal as it is digitized and calculating the root mean square (RMS) of the voltage. The RMS value is calculated over consecutive time intervals of .256 second (ie. 256 digitized words) for the duration of data aquisition.

A typical example of the original signals obtained from one of the subjects performing task 1 is shown in Fig. 2-5. The corresponding RMS value is plotted versus time in Fig. 2-6. The strain gauge output is superimposed on the RMS output to illustrate the correlation between the rise in the EMG output with the subject's extensor effort.

The sample results show that the spinalis capitis activity increases with the extensor effort from the outset. Increasing muscle activity with increasing extensor effort is not observed in the sternomastoid, omohyoid, and splenius capitis muscles until a much greater extensor effort is reached. It is noteworthy that for this particular example the onset of muscle activity for the right and left sides of the splenius capitis do not occur for the same value of extensor effort. This phenomena is also very pronounced in the omohyoid muscle.

Variations are observed in the results between the different volunteers, between tests performed on the same volunteer (as illustrated by the sample results of Fig. 2-6) and between results from the muscle mass on opposite sides of the spine for the same muscle during the same test. It is felt that the results for each of the 5 tasks could be best interpreted as the average of the three tests performed on each of the volunteers and as the average of all the tests on all of the volunteers.

A summary of these results are shown in Table 2-2

TASK	VOLUNTEER	MAXIMUM EFFORT (kg)	MUSCLE ACTIVITY ONSET STERNOMASTOID (kg)	OMOHYOID (kg)	SPINALIS CAP (kg)	SPLENIUS CAP (kg)
1	1	25.8	10.1	16.0	0.0	5.7
	2	25.4	13.2	12.5	2.4	7.5
	3	17.7	7.3	11.9	0.0	6.7
	4	19.0	5.4	9.3	0.0	4.6
	5	26.2	15.1	19.8	0.2	7.5
	average	22.8	10.2	13.9	0.5	6.4
2	1	22.5	5.3	9.7	1.1	3.0
	2	20.0	14.1	16.1	3.6	4.8
	3	18.6	7.9	10.4	1.2	0.7
	4	19.2	8.0	12.0	0.3	3.1
	5	18.3	8.8	13.3	3.0	4.4
	average	19.7	8.8	12.2	1.8	3.2
3	1	30.4	8.2	11.2	0.0	8.5
	2	27.0	8.0	18.8	0.4	10.9
	3	21.5	14.5	14.4	0.0	11.1
	4	19.7	6.6	6.0	0.0	4.1
	5	30.4	9.1	22.4	1.4	9.1
	average	25.8	9.3	14.6	0.4	8.7
4	1	20.1	9.5	13.0	2.7	4.2
	2	17.8	6.6	12.4	0.6	5.1
	3	16.8	21.1	14.3	4.1	5.4
	4	20.6	7.4	10.8	0.0	1.6
	5	23.3	12.0	15.6	0.8	3.8
	average	19.7	11.3	13.2	1.7	4.0
5	1	32.0	10.2	14.0	0.0	7.0
	2	29.3	16.3	16.3	0.0	9.0
	3	23.5	9.5	11.4	0.9	8.3
	4	25.3	7.4	11.6	0.3	4.3
	5	27.4	12.6	14.4	0.2	6.1
	average	27.5	11.2	13.5	0.3	7.0

Table 2-2 Averaged results obtained from volunteers performing TASK 1 to 5

2.4 DISCUSSION

In the numerical description on the cervical spine (appendix A) each muscle is described as a collection of vectors running from points of origin to points of insertion. Each group can be approximated as a single vector representing the muscle's average line of action (Fig. 2-7). This approximate representation is useful in interpreting these experimental results.

Semispinalis Capitis and Splenius Capitis: The findings of the present study indicate that as the extensor moment is increased, the spinalis and semispinalis capitis and splenius capitis muscles are recruited in two distinctly different ways. As the extensor moment is increased, the spinalis and semispinalis capitis muscle exhibits an immediate response whereas the splenius capitis does not. This delay is observed to be a function of extensor moment (ie. applied load). The only previous report on the function of these muscles is that of Takebe, Vitti, and Basmajian [44] which showed the EMG activity of these muscles versus time for free motion; the above phenomenon is not observed, since muscle activity as a function of the extensor effort is not recorded.

Sternohyoid and Omohyoid: These two muscles produce flexion moments on all of the intervertebral cervical joints. Since the neck is producing an extensor effort, it is reasonable to assume that these muscles would produce no activity during the execution of this task. This intuitive assumption is observed in both the preliminary and main investigation for the sternohyoid muscle but not for the omohyoid muscle.

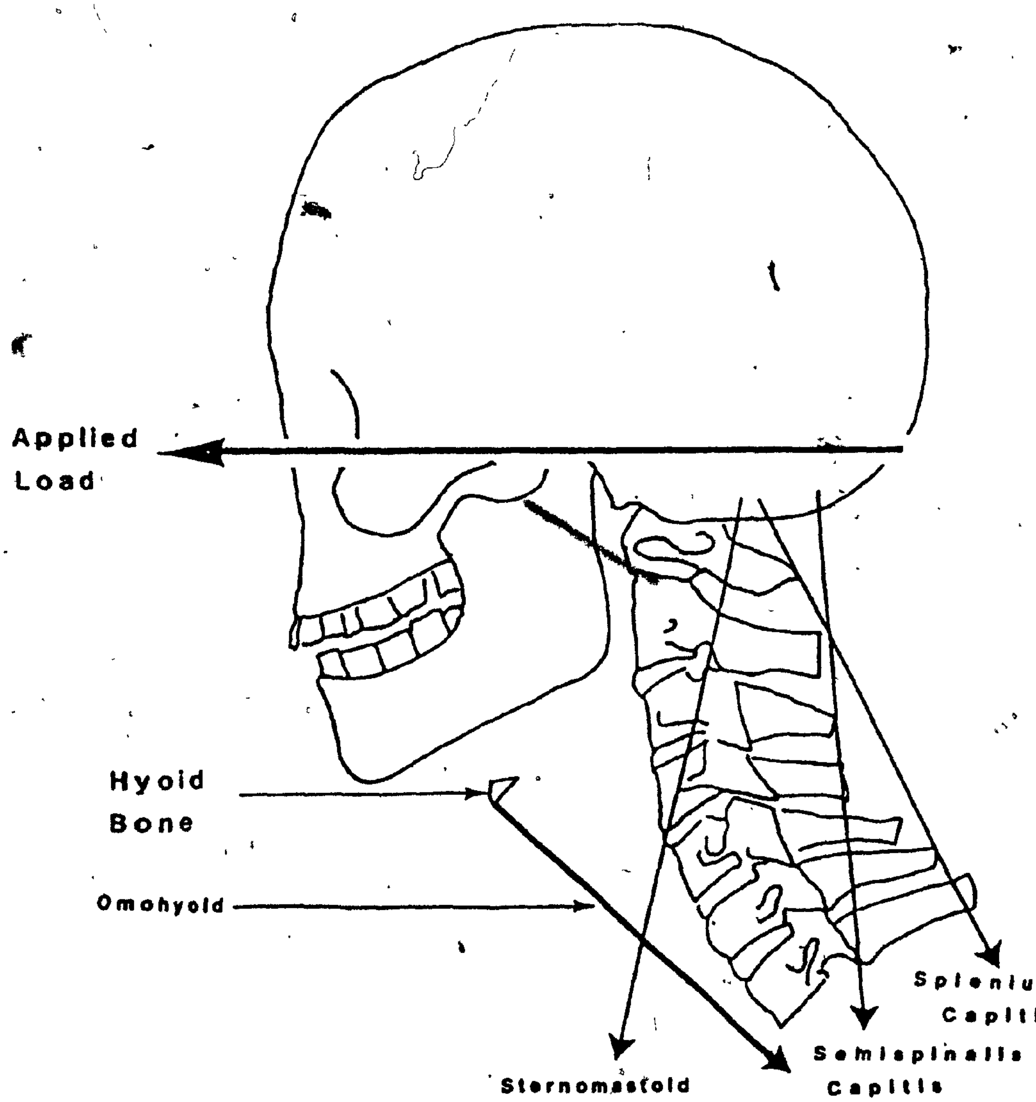


Figure 2-7 Schematic average force produced by muscles of interest

This study shows the omohyoid muscle to be active during extension of the head against resistance. Because this muscle is an anterior muscle and is usually considered a flexor of the head, it is surprising to find this muscle active during extension of the head. This fact makes this finding especially significant. Furthermore, if the omohyoid is indeed working at a significant portion of its assumed maximum during resisted extension, it would mean the discovery of a new secondary function for the muscle.

Integrated EMG results from the omohyoid show a wide spread of values. It is not always possible to palpate this small muscle and hence ease of electrode placement varied considerably from subject to subject. This may account for most of the scatter. If so, a careful screening of participants for accessible omohyoids may result in a more accurate study.

Intuitively, one can see from Fig.2-7 that the extensor effort will produce relatively larger flexion moments at the lower cervical joints than at the higher cervical joints. Therefore in order to balance this applied moment it is necessary for the muscle to produce a relatively higher extensor moment at the lower cervical joint than at the upper ones.

The capitis muscles tend to produce a large moment at the upper joints as well as at the lower ones. If use is to be made of these muscles in the execution of this task, it is necessary to recruit another muscle which can reduce this undesirable effect of the capitis muscle in the upper joints. The omohyoid has an orientation that is well suited to counterbalance this moment. Although its diameter is small, it has a long lever arm. For this reason it should not be overlooked as a substantial contributor toward stabilization of the cervical spine during the performance of this task.

Sternomastoid and Scalene Posterior: A recent investigation of the sternomastoid [44] has shown this muscle to be rarely active during backward extension of the head against resistance. These researchers have reported activity during this exercise in only 4 out of 20 subjects examined. In the preliminary investigation of the present study activity is noted in all but one case. The results for this muscle can be considered more statistically reliable than in the case of the omohyoid due to the ease of electrode placement on the sternomastoid muscle.

From Fig. 2-7 it can be seen that this muscle produces a large flexor moment at the lower cervical joints. Activity from this muscle is thought to be undesirable at a time when priority has been given to balancing the very large flexor moments resulting from the load at these lower joints. For the same reason, no activity is expected from the scalene muscles and the sternohyoid muscle. It is not known why a slight degree of activity from the sternomastoid muscle is observed.

Similar results can also be observed from the posterior portion of the scalene muscle. This muscle produces almost pure compression at all of the cervical joints with insignificant extension or flexion moments. Since this muscle is not in a position to support the load, it is difficult to explain the role of this muscle on an intuitive basis. It is possible that the scalene and sternomastoid muscles are acting to support the rib cage rather than the neck.

CHAPTER 3 MODELLING THE HUMAN CERVICAL SPINE

3.1 OBJECTIVE

The purpose of this section is to develop a sagittal plane mathematical model capable of simulating the mechanism of the musculo-skeletal structures of the neck when subjected to the forces resulting from sagittal plane loading conditions such as pilot ejection.

Examination of previous modelling approaches reveals that the continuum modelling approach is incapable of determining the specific structure which would limit the spine in supporting a load. The discrete modelling approach is forced to rely on assumed parameters when insufficient test data is available. A greater problem of the discrete parameter model is its inability to determine the contribution of the muscles. This second limitation is perhaps the most serious since it is felt that the muscle will play a very critical role in the supporting of a load or the execution of a movement.

It is felt that the best approach to follow in developing the required model is the optimization approach used by Gracovetsky et al [18] in modelling the lumbar spine. This approach is particularly attractive since it makes possible the use of EMG results in tuning the model. Therefore the same principles which are used in the development of the lumbar spine model are to be used to develop a model of the cervical spine.

3.2 THE MODEL

When an external load is applied to the spine, the moment it creates at each intervertebral joint must be balanced by internal moments created by muscle and ligament tensions. To illustrate how this balance is achieved, a single cervical joint, as shown in Fig. 3-1, is considered.

In this illustration, it can be seen that the joint supports the weight of the head and that portion of the neck above it. In order to simplify the diagram further, only a single muscle strand of semispinalis cervicis and a single strand of splenius cervicis are shown.

The forces acting on this cervical joint may be divided into 4 groups:

1)- The load, resulting from the head and neck

2)- The reaction forces of the joint, acting at the center of reaction

3)- The muscle tensions, shown here to be acting behind the center of reaction.

4)- The ligament tensions, shown to be acting behind the center of reaction.

In order for equilibrium to be achieved the sum of the components of all the forces in the two mains directions (i.e shear direction and compression direction) must sum up to zero and the moments must balance.

The reaction force can be seen as a fulcrum around which the moment due to muscle and ligament tensions balance the moment due to the load.

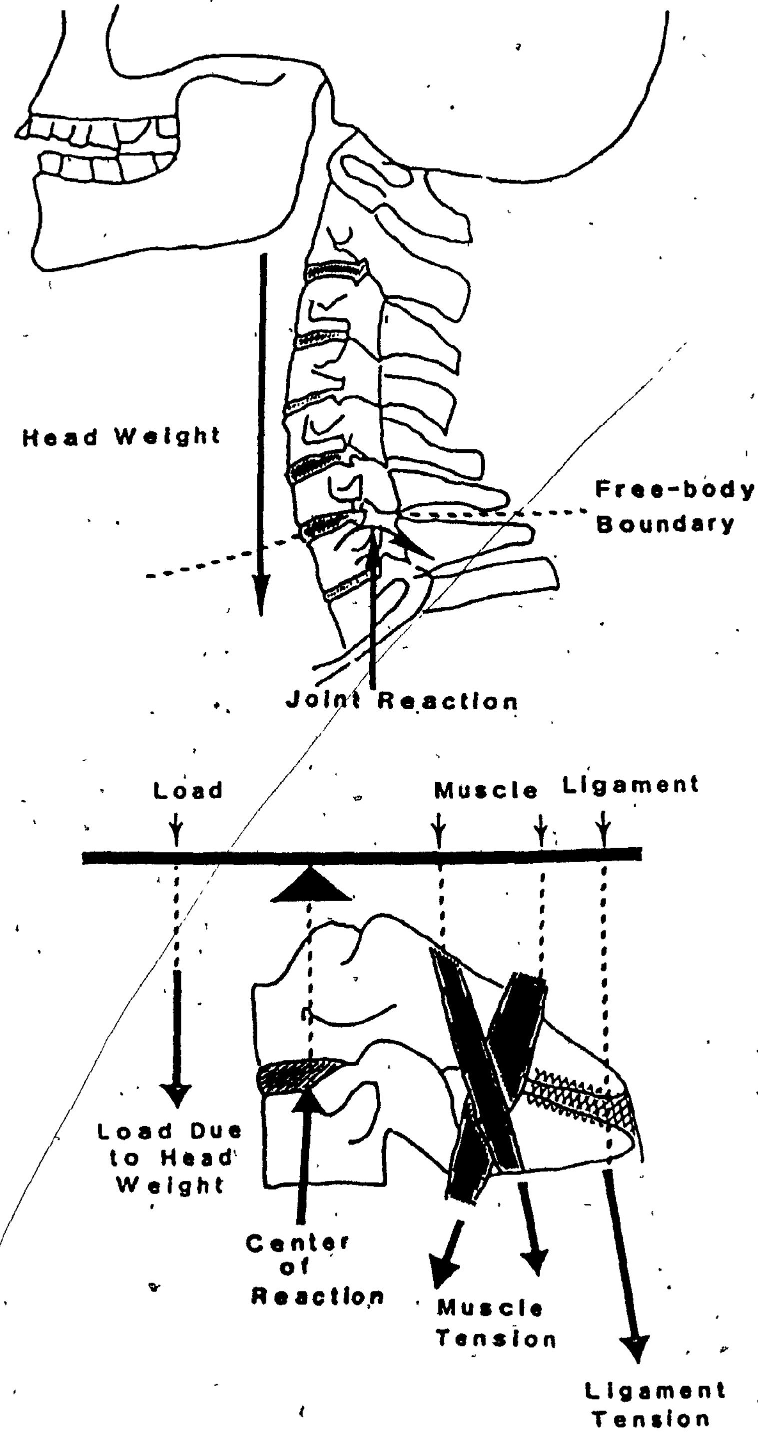


Figure 3-1 Simplified free-body analysis of C6-C7 intervertebral joint showing load distribution in a cervical joint

From the diagram it becomes obvious that the load can be balanced by an infinite number of combinations of muscle and ligament tensions. Hence it is necessary to introduce additional constraints on this system in order for the equilibrium conditions problem to be a determinate one.

Wolff's law, as stated elsewhere in this report, characterizes the response of the bone to stress. It is proposed that Wolff's law can be mathematically expressed by the fact that stress within the spine must be minimized and equalized. Therefore, it is reasonable to search for the muscle action that will not only balance the load, but will also tend to produce stress minimization and equalization at each intervertebral joint.

The problem is mathematically formulated as the minimization of an objective function subject to various types of constraints.

3.2.1 The Objective Function

The objective function expresses the necessity for the intervertebral joints to minimize and equalize their stress. In this way optimum use of the muscular power and the strength of ligaments is obtained. The stress is due to the actions of the muscle pull (K_k), the ligament reaction (L_{ij}) and the shear and compression reactions of the joints (J_{ij}). Where k is the muscle group from 1 to the number of groups (Nm), i is the force component (ie. shear and compression) and j is the joint level from 1 to 8. This objective can be approximated as a quadratic function as follows:

$$F = F_1(\text{muscle}) + F_2(\text{shear}) + F_3(\text{comp.}) + F_4(\text{ligament}) \qquad (3)$$

where:

$$F_1(\text{muscles}) = \sum_{k=1}^{Nm} (P_1 x K_k)^2 \quad (3a)$$

$$F_2(\text{shear}) = \sum_{j=1}^{8} (P_2 x J_{1j})^2 \quad (3b)$$

$$F_3(\text{comp.}) = \sum_{j=1}^{8} (P_3 x J_{2j})^2 \quad (3c)$$

$$F_4(\text{ligament}) = \sum_{j=1}^{8} (P_4 x L_{ij})^2 \quad (3d)$$

This form of objective function is chosen for two reasons:

1) it adequately represent physical objective of stress minimization and equalization

2) it is reasonably straight forward to obtain a mathematical solution

The P_1, P_2, P_3 and P_4 are, at this point, constant coefficients determining the relative importance of the various terms in the objective function. These coefficients represent the only freedom that this modelling has, and all experiments must be accounted for by the determination of these four quantities.

The coefficients P_1, P_2 and P_3 are set respectively to be the inverse of the maximum muscle pull, the joint shear and the joint compression. In this way the muscle and joint stresses will tend to achieve their maximum value for equal values of F_i.

At this point, only P_4 needs to be determined. This can be done by imposing the requirement that the optimized model output show reasonable agreement with the experimental results.

It should be noted that P_4 is intimately attached to the concept of muscular strategy in the sense that the task may be performed in a number of ways depending on how the subject chooses to recruit his ligament structure.

3.2.2 Equality Constraints

The solution of the problem must satisfy the equations of equilibrium. These are also called the equality constraints. The forces acting at a joint can be considered to be due to one of four groups:

1) EXTERNAL LOAD - defined as a force due to the weight of head and neck together with an externally applied load.

2) MUSCLE LOAD - defined as a force due to the muscles when contracted.

3) LIGAMENT REACTION - defined as a force due to the deformation of the ligament, as a result of motion.

4) JOINT REACTION - defined as a force due to the deformation of the disc or the facet.

Since there are no other forces acting on the joint, the moments, shear, and compression due to these forces must balance. This can be stated in equation form as follows:

$$\sum_{k=1}^{Nm} (A_{ijk} \times K_k) + E_{ij} + L_{ij} + J_{ij} = 0 \qquad (4)$$

Where:

A_{ijk}=force component(i) (compression, shear, and moment) at joint(j) (C0-C1, C1-C2,...,C7-T1) due to a unit stress in muscle group k (multifidus,scalene,etc.)

K_k =stress in muscle group k

E_{ij} =force component i at joint j due to an external load

L_{ij} =component i at joint j due to the reaction of the ligaments at joint j

J_{ij} =force component i of the reaction at joint j due to the disc and the facet

The A_{ijk} terms can be determined from anatomical descriptions of the cervical spine and x-rays (Appendix A). The values for E_{ij} can be obtained from a precise description of the task to be performed.

The mathematical objective is to determine the muscular action (K_k), the ligament tensions (L_{ij}), and the joint reactions (J_{ij}) due to an external load E_{ij} with the motion of the neck restricted to the sagittal plane.

3.2.3 Inequality Constraints

With the convention that positive stress represents tension, the inequality constraints may be expressed as follows:

$$K_k \geq 0 \qquad (5a) \qquad \text{and} \qquad L_{ij} \geq 0 \qquad (5b)$$

That is to say that the muscles and ligaments can only exert a pull. The modelling procedure then consists of minimizing the objective function ($F = F_1 + F_2 + F_3 + F_4$) in such a way that the equality and inequality constraints described above are satisfied.

3.2.4 Determination of the Coefficients P_1, P_2, P_3 and P_4.

As stated earlier, the coefficients are set to homogenize the stress values in the structural components of the neck. An appropriate method for choosing these values is to set them equal to the inverse of the square of the maximum stress that can be withstood in the corresponding structural group. Therefore as an initial estimate for P_1, P_2 and P_3, one chooses the best available estimate for the maximum values of the muscle pull, the joint shear and the compression. The initial approximations taken are:

$$P_1 = 1. \qquad P_2 = .01 \qquad P_3 = .01$$

More accurate estimates of these values will be determined by trial and error using experimental data. Since the ratio of the coefficient is what determines the optimal solution of the minimization problem, the coefficients are normalized in such a way that P_1=1.

The significance attached to the ligament weighting factor P_4 is different. The parameter P_4 ensures that the subject cannot arbitrarily recruit his ligament structure in performing the task. If, for example, the subject is required to support the load in a neutral position, one would expect that the ligaments would produce very little tension. This would be equivalent to assigning a relatively high value to the parameter P_4.

If the subject is required to support a load while in a flexed position, it would be reasonable to expect the ligaments to support a much greater portion of the load. This would be equivalent to setting the parameter P_4 to a much lower value. The determination of P_4 is done by tuning the calculated muscle response to the measured EMG response.

3.2.5 Reduction of model size: muscle grouping.

The numerical description of the neck requires every muscle to be described as a vector running from its point of origin to its point of insertion. In theory, it is possible to consider each of these vectors (muscles) as independent variables; but this would result in an unreasonably large problem. It is therefore desirable to put forward some simplifying assumptions. Such simplification exists due to the fact that the motion has been restricted to lie in the sagittal plane. As such each muscle can not behave independently. In other words, the muscle strands are expected to fire in functional groups. Determining the minimum number of independent muscular groups is therefore an important step since it results in a significant reduction in the computational burden necessary for solving the problem.

The determination of which muscle strands belong to which group is a trade-off between reducing the number of variables to a manageable level without sacrificing the freedom necessary to execute the task. Therefore it is importartant to state the properties that a muscular group can or cannot possess in the execution of the task:

PROPERTY # 1 : All the muscle strands in a group have their activity increasing or decreasing at the same time. An example of an action of the semispinalis capitis would be extension of the head and neck against resistance. As the extensor effort increases, so does the activity of all the muscle strands.

PROPERTY # 2 : Muscles which traverse completely different sets of joints will have different actions and therefore will be assigned to different groups. For example, consider the splenius cervicis which inserts into the neck and the splenius capitis which inserts into the head. These two muscles are treated as independent variables, and will be assigned to different groups. Similar considerations apply to the semispinalis cervicis and capitis, the longissimus cervicis and capitis, and the longus cervicis and capitis.

PROPERTY # 3 : Muscles with completely different functions will have different actions and should be placed in separate groups. Examples of this are the omohyoid which produces a flexor moment, the scalene muscles which produce little moment, and the splenius cervicis which produces an extensor moment.

Using these assumptions and some trial and error a muscle grouping arrangement is obtained. The resulting muscle groupings are shown in Table 3-1 below.

Classification of muscle activity For any given task, the muscular action can be characterized as follows:

1- Primary Action- A muscle having a primary action will perform a task at near maximum level by firing at near maximum level.

2- Secondary Action- A muscle having a secondary action will perform a task at near maximum level by firing at a level inferior to its capacity.

Group	Muscle	Group	Muscle
1	Multifidus (C2-C3 level)	14	Longus Capitis
2	Multifidus (C3-C4 level)		Longus Superior
3	Multifidus (C4-C5 level)		Longus Vertical
4	Multifidus (C5-C6 level)		Longus Inferior
5	Multifidus (C6-C7 level)	15	Scalene Posterior
6	Multifidus (C7-T1 level)		Scalene Medius
7	Semispinalis Cervicis		Scalene Anterior
8	Semispinalis Capitis	16	Sternohyoid
9	Spinalis capitis	17	Sternomastoid
10	Splenius Capitis	18	Omohyoid
11	Splenius Cervicis	19	Rectus Capitis Minor
12	Longissimus Capitis	20	Rectus Capitis Major
13	Longissimus Cervicis	21	Oblique Capitis Superior
	Iliocostalis	22	Oblique Capitis Inferior

Table 3.1 Grouping of muscle vectors into functional groups

3.3 MODEL TUNING AND SIMULATION

Tuning is the procedure by which the model response is made to match the experimental EMG data obtained from human experiments. This adjustment is made by appropriate modification of the parameters P_2, P_3 and P_4.

3.3.1 Tuning of the parameter P_4

Increasing the value of P_4 has the effect of forcing the model to rely on muscular action to support the load. The values of P_2 and P_3 are set to the initial value of .01 as mentioned earlier. P_4 is initially set to zero and gradually increased until the muscles reach their maximum for the maximum extensor effort obtained in the experimental investigation (Chapter 2).

The maximum force per unit cross-sectional area that the muscles can produce is about 8 kg/cm^2[20]. Based on the study of Gracovetsky, Farfan, and Lamy [17], it is assumed that the maximum voluntary muscular effort performed by the volunteers will not exceed 2/3 of this ultimate limit.

With this assumption the following values of P_4 for the flexed, upright and extended neck postures are respectively 0.04,0.09 and 0.10.

Initially, the experimental investigation of Chapter 2 is executed with the semispinalis capitis and the spinalis capitis grouped together. This grouping resulted in calculated responses that are not confirmed by EMG experiments. Hence, these two muscles are split into two distinct muscle groups. This arrangement has been confirmed by EMG data for the three postures and is adopted in all subsequent simulations.

3.3.2 Tuning of the Parameter P_2

The value of the parameter P_2 has been increased and decreased around its initial guess to determine the consequences on the load distribution in the neck. Changes in P_2 do not greatly change the distribution of the load between muscles and ligament. It does affect however the distribution of the load between the muscles themselves, notably the spinalis capitis and the splenius capitis.

It is important to remember that at this stage the three parameters P_2 to P_4 represents the only degrees of freedom available in the model to tune the response of the eight intervertebral joints to a load. This arrangement proved inadequate for the following reason:

A single value for P_2 for all eight intervertebral joints resulted in poor correlation between the simulation and experiments. With a value of $P_2 = 0.01$, the onset of the calculated splenius capitis activity is in the range of 0 to 1kg . By increasing P_2, little change in the model response is observed. By decreasing P_2, the onset of the calculated splenius capitis activity would tend towards 0 kg. These results do not correspond well with the measured onset of activity (5 to 6kg range).

As a next step, a separate weighting coefficient is assigned to the upper two cervical joints (occipital-C1-C2). By augmenting the value of this parameter, it is noted that a significant increase in the extensor effort required to recruit both the splenius capitis and the omohyoid. The response of the model remains insensitive to changes of the P_2 value corresponding to the lower joint due to the domination of the model's response by the shearing load on the upper two intervertebral joints.

This result suggests that the resultant of any force vector passing through the occipital-C1-C2 region must contain minimal shear component. The entire cervical system will react strongly to the presence of any shear in this area.

The best values for the parameter P_2 are 8.0 for the upper two cervical joints and 0.03 for the lower 6 cervical joints.

3.3.3 Tuning of the Parameter P_3

The parameter P_3 controls essentially the compression at the joints. By increasing P_3 above the initial value of 0.01 it is noted that a shift in the load distribution from the muscles with shorter lever arms, such as the multifidus, to the muscles with longer lever arms, such as the semispinalis cervicis and spinalis capitis. Large increases in P_3 also resulted in the elimination of the omohyoid's contribution.

On the other hand, by decreasing the value of P_3 it is noted that there is little change in the response of the model. This suggests that the lower values of P_3 may be correct, since they indicate a higher compressive strength. A value of 0.01 is used in all subsequent simulation since it represented the most conservative estimate.

3.3.4 Simulation of experimental tasks

Using the weighting coefficients selected above, the model is used to simulate the five tasks that the volunteers performed in the experimental investigation (Chapter 2). The line of action of the resistive force of the restraining strap can be obtained from photographs of the subject extending his neck against resistance. The results of these simulations are described below and illustrated in Fig. 3-2.

TASK # 1 The neck is in the normal upright posture. The average height of the restraining strap is 21cm above the subject's sternum and the average angle between the restraining strap and the horizontal is 10 degrees. The onsets of muscle activity for the spinalis capitis, the splenius capitis and the omohyoid are respectively 0, 4, and 15 Kg.

TASK # 2 The neck is in the fully flexed posture. The restraining strap is at an average height of 22cm above the sternum and the average angle with the horizontal is 13 degrees. The onsets of muscle activity for the spinalis capitis, the splenius capitis and the omohyoid are respectively 0, 0 and 9 kg.

TASK # 3 The neck is in the fully flexed posture. The restraining strap is at an average height of 17cm above the sternum and the average angle with the horizontal is 11 degrees. The onset of muscle activity for the spinalis capitis is 3kg. The splenius capitis and omohyoid are not active.

TASK # 4 The neck is in the fully flexed posture. The restraining strap is at an average height of 25cm above the sternum and the average angle with the horizontal is 45 degrees. The onset of muscle activity for the spinalis capitis, the splenius capitis, and the omohyoid is 0 kg.

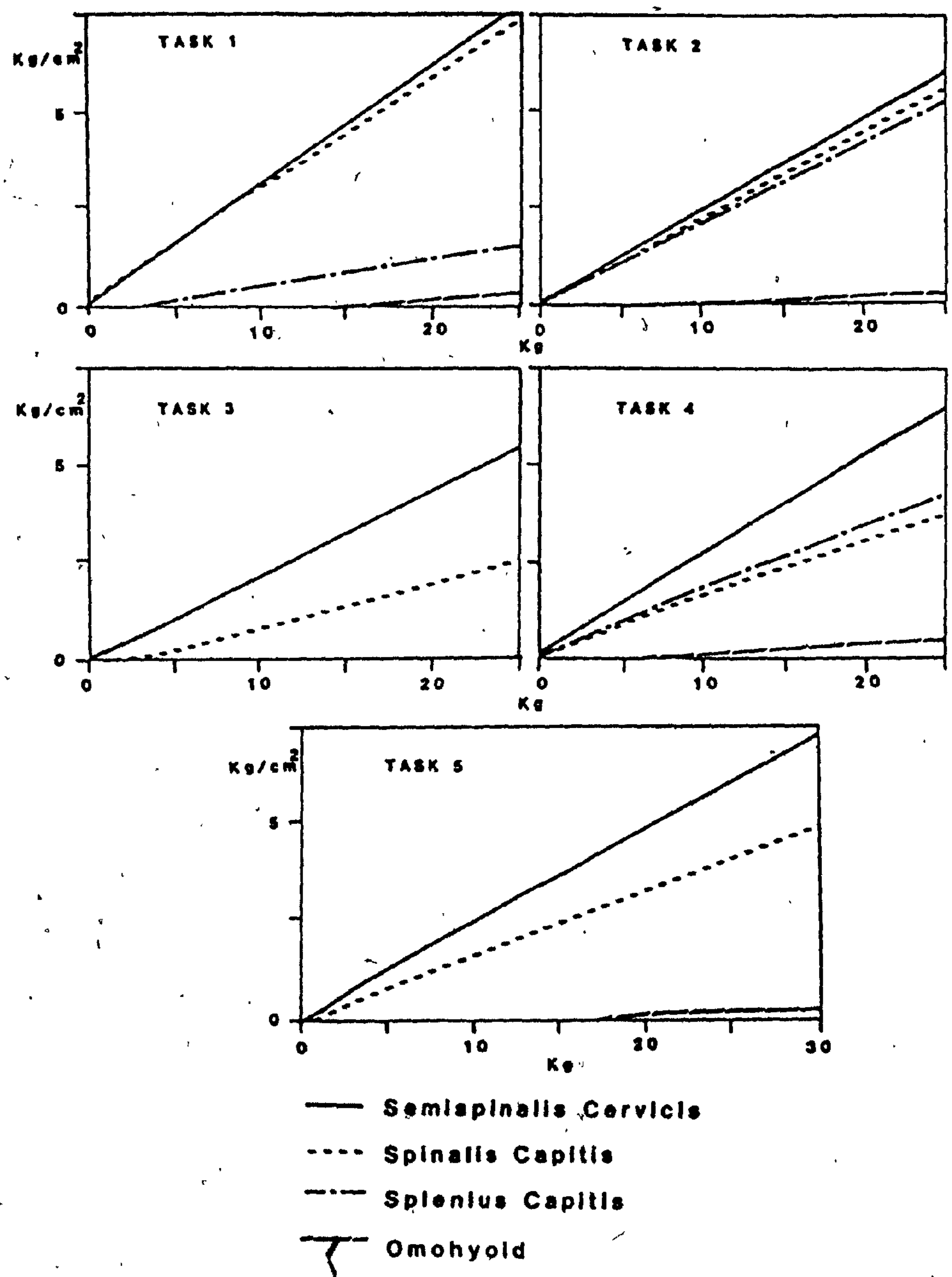


Figure 3-2 Simulation results for neck subjected to loading resulting from subject executing task 1-5

TASK # 5 The neck is in the normal upright posture. The restraining strap is at an average height of 19cm above the sternum and the average angle with the horizontal is 41 degrees. The onsets of muscle activity for the spinalis capitis and the splenius capitis are respectively 0 and 14kg. The omohyoid is not recruited.

3.4 DISCUSSION

The five tasks have been simulated with the same values of P_2 and P_3. The parameter P_4 is a function of the posture and is varied depending on the task. The model results are compared with the average experimental results of Table 2-2.

3.4.1 Relation to EMG Results

During the switch from a flexed posture to an extended posture, the level of extensor effort required to recruit the splenius capitis and the omohyoid increases. When the strap position is lowered to produce a larger angle with respect to the horizontal, these experimental results indicate a slight increase in the extensor effort required to recruit the splenius capitis muscle.

The same trends are observed in the simulated results but the calculated trends appear to be exaggerated. One possible explanation is that the model is based on X-rays of an unloaded individual extending and flexing his head to his limit. The same individual under loaded conditions can be expected to perform the same task at a slightly different angle of flexion. It is not desirable to X-ray volunteers for such determination. Also the restraining strap represents an additional constraint on the attitude of the subject's head while performing the tests.

It is also noted that the calculated muscular pattern results from an optimization procedure. Such a procedure results in a unique mathematical solution. The variations in the muscular response from the various volunteers and even the variations observed between tests on the same volunteer cannot be expected to be precisely accounted for with such a deterministic approach.

3.4.2 Relation to Physiological Behaviour

In the accomplishment of any given task, the spine follows basic laws of physics resulting in measurable physiological behavior (i.e. specific muscular pattern, a specific geometry, a specific disk pressure). The mathematical equations minimizing stress yield muscle patterns specific to the task. Using the muscle patterns of only two groups of muscles, an EMG pattern is calculated for 22 muscle groups. The EMG pattern of all muscles available to surface electrodes are found to conform to the calculated results. This is true for a range of neck postures. Striking in this regard is the predicted function of the omohyoid and its completely unsuspected role as a flexor of the head.

There are two exceptions to this. The sternomastoid exhibits low level activity which is attributed to a possible small positional shift of the restraining strap. There is also low level activity in the scalenes which is believed to be related to the fixation of the chest with their muscular effort.

At the present time there are no measurements in the literature of disc pressure in the cervical spine. The values obtained from the simulation are well within known limits for lumbar discs.

Although it cannot be concluded that physiological behavior has been truly represented, it is felt that the system of loading in the spine has been closely approximated . This approximation appears to be the same in all major respects to the physiological system of loading of the lumbar spine.

CHAPTER 4 SIMULATION OF NECK RESPONSE TO ACCELERATION LOADING

The next stage of the study is to use the tuned model to simulate a subject undergoing high constant acceleration. For this study there are four possible parameters which can be varied. The first is the magnitude of the acceleration which the subject must sustain. The second is the direction of acceleration with respect to the subject. Finally, changes in the subjects posture must also be considered.

The problem of using the model to obtain conservative estimates of the maximum acceleration which can be supported by the neck is approached in two ways. The first is the case in which the moments generated by the acceleration load is supported only by the musculature (MUSCLE STRATEGY). These results are independant of the experimental results. This serves as a worst case analysis of how much acceleration the neck can support.

For the second approach, the ligaments are introduced as significant moment supporting structures (LIGAMENT STRATEGY) in order to obtain a more realistic analysis of the cervical spine's load bearing capacity. The role of the ligaments is estimated with the use of the experimental results obtained in Chapter 2 and the results of the simulation of these experiments obtined in Chapter 3.

4.1 MUSCLE STRATEGY

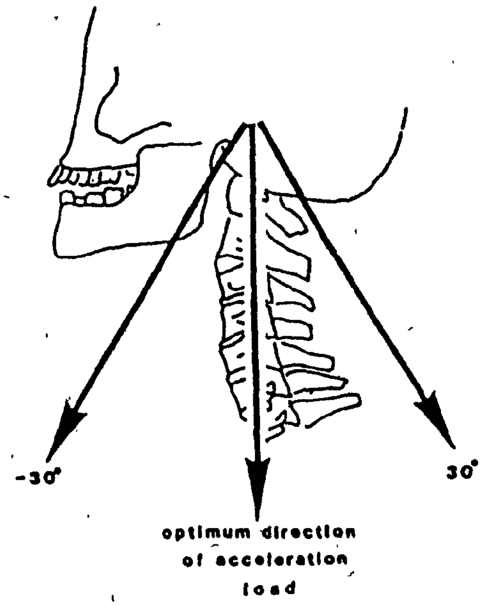
The model is tuned in such a way that the ligament tensions are as low as possible while still maintaining equilibrium at the cervical joints. This is achieved by setting the ligament weighting factor P_4 to as large a value as possible without altering the muscle firing strategy by an unreasonable amount. A value of $P_4 = 1.0$ is used for this purpose. The remaining weighting factors P_2 and P_3 are set to the initial guesses discussed in section 3.2.4 of .01 and .01. In this way dependancy on the experimental results is avoided.

Results are calculated for the normal upright (or neutral), flexed and extended postures. Acceleration loads producing flexion moments and extension moments are considered.

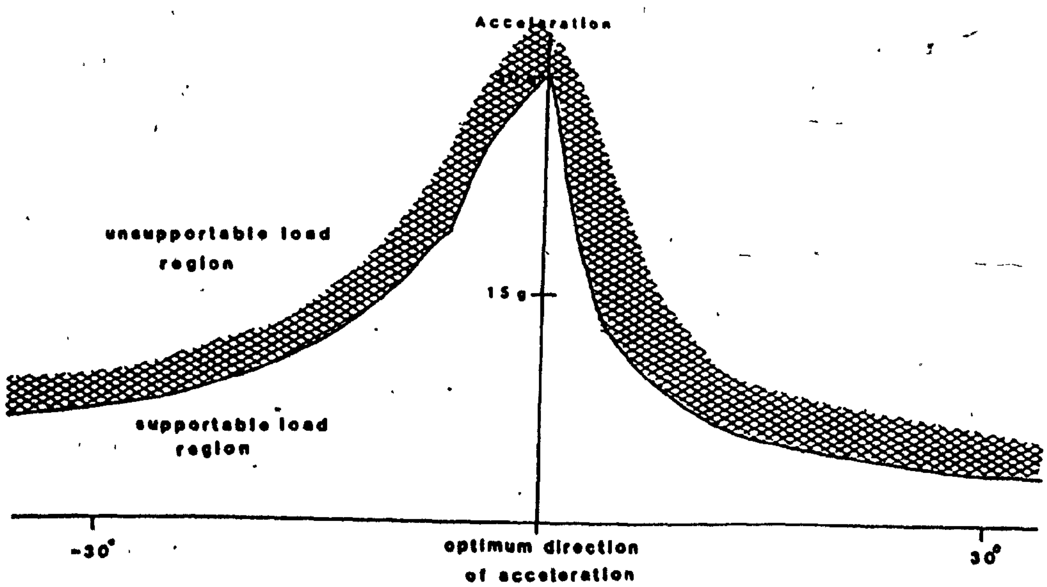
4.1.1 Neutral Geometry

For the first set of simulations the neck is represented in the normal upright posture (neutral geometry). The model is used to obtain the muscle, the ligament and the joint responses to this acceleration. The value of the acceleration load is increased until one of the structural components reached its limit.

The relative arrangement of acceleration vectors with respect to the cervical spine is altered as indicated in Fig 4.1a. The maximum acceleration supportable for each direction of the acceleration vector is calculated in accordance with the above procedure. The results are shown in Fig. 4-1b as a plot of the maximum acceleration that the muscles of the neck could support versus the orientation of the line of acceleration with respect to a line acting through the center of gravity of the head and approximately through the axis of the spine.



a) loading configuration



b) results from simulation

Figure 4-1 Maximum acceleration which can be supported by the muscles with the cervical spine in the neutral position

It can be seen that if the subject assumes a neutral spinal geometry, he can support up to 30g of acceleration when the load is acting through the spine. If however the acceleration deviates slightly from this optimum orientation, the maximum acceleration that the muscles may support decreases.

4.1.2 Flexed Geometry

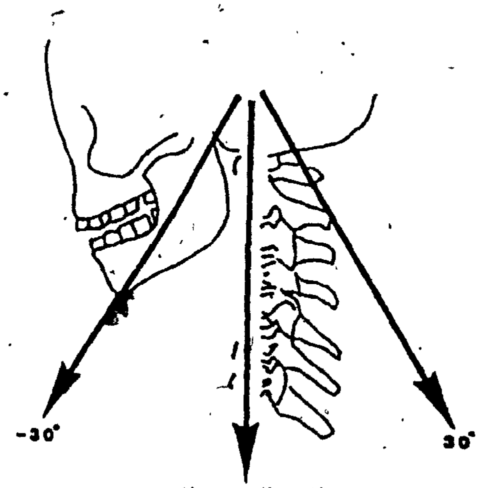
This simulation considers the spine to be in a flexed geometry. As is done in the previous case, the acceleration is increased until one of the muscles of the neck reached its limit.

The orientation of the line of acceleration with respect to a line acting through the center of gravity of the head and approximately through the axis of the spine is varied (Fig. 4-2a). Values of maximum acceleration which the muscles could support are determined. These results are presented in Fig. 4-2b as a plot of the maximum supportable acceleration versus orientation of the line of acceleration.

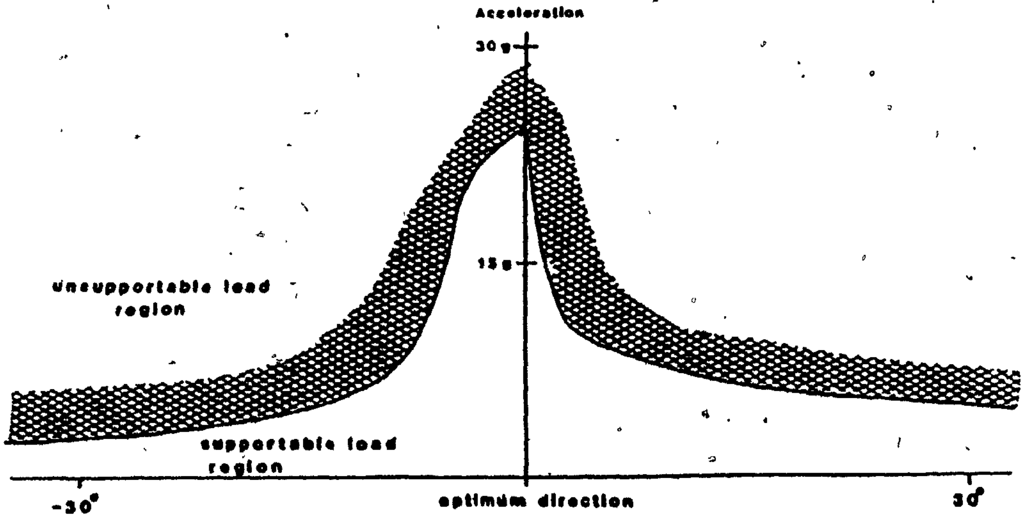
It can be seen that the maximum acceleration that can be supported in the flexed spinal geometry occurs when the line of acceleration is acting approximately through the spine. The important feature is that the maximum acceleration dropped from 30g for the neutral position to 24g for the flexed position. As seen in the previous results the maximum decreases sharply for change in the orientation of the subject.

4.1.3 Extended Geometry

Finally simulations are performed for the neck in the extended spinal geometry. As is done in the previous two cases, the acceleration is increased until one of the cervical muscles reached its limit.

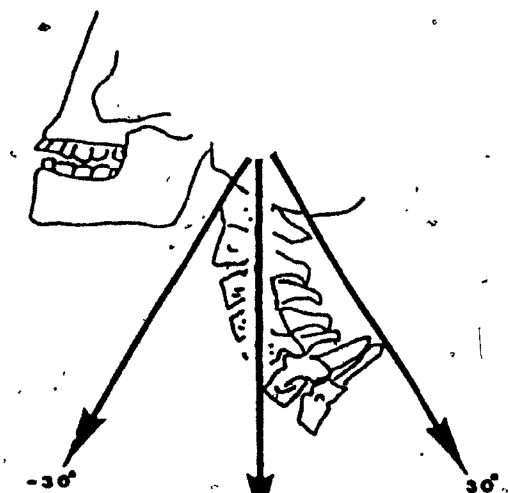


a) loading configuration

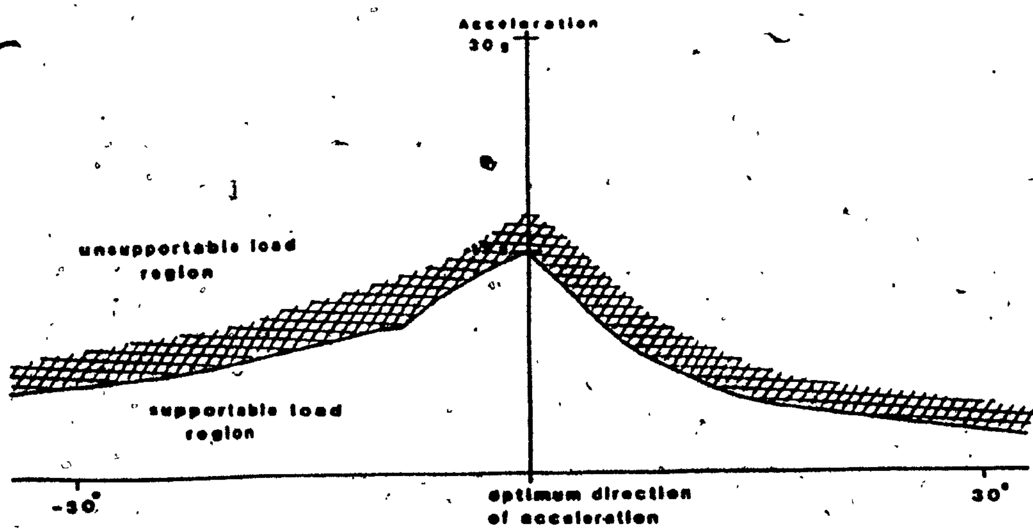


b) results from simulation

Figure 4-2 Maximum acceleration which can be supported by the muscles with the cervical spine in the flexed position



a) loading configuration



b) results from simulation

Figure 4-3 Maximum acceleration which can be supported by the muscles with the cervical spine in the extended position

The orientation of the line of acceleration with respect to a line acting through the center of gravity of the head and approximately through the axis of the spine is varied (Fig. 4-3a). Values of maximum acceleration are determined and the results are presented in Fig.4-3b.

It can be seen that the maximum acceleration that can be supported in the extended spinal geometry occurs when the line of acceleration is acting approximately through the spine. The important feature is that the maximum acceleration dropped from 30g for the neutral position to 15g for the flexed position.

4.2 LIGAMENT STRATEGY

In order to obtain a more realistic simulation of the response of the cervical spine to acceleration load, it is necessary to introduce the ligaments as a structure which supports a large portion of the moment resulting from the load. Only accelerations producing flexor moments can be considered due to the nature of the idealization of the ligaments used in the model.

Two types of acceleration loads will be considered to illustrate the model response. In the first case, the acceleration direction will be determined by the condition that the resultant load at each joint be essentially compressive. This will result in a small flexor moment at the joints. The second case will be represented by an acceleration direction perpendicular to the previous acceleration direction. This will result in a very large flexor moment at the joints.

Acceleration directions which produce extension monents are not considered. A large extension moment would require that the anterior ligaments support a significant portion of the load. Since experimental results are obtained for a flexion load, the extension of those experimental results to extension type loading cannot be justified.

For each of these two acceleration directions, two neck postures are considered; a normal upright posture and a fully flexed posture. Hence, a total of four cases are analyzed. The extended neck posture is not being considered in this set of simulations since acceleration directions producing extension moments cannot be considered and it is difficult to conceive of an ejection sequence which would result in a flexion moment with the head in the extended posture.

For each of these four cases, it is desired to determine the maximum supportable acceleration. Of course this maximum value depends upon the maximum muscular firing density. This is not known, but the maximum voluntary effort can be estimated by using simulation results of the experimental tasks presented earlier. This analysis is based on two assumptions:

> The first is that when the volunteer reaches his maximum voluntary extensor effort, he also reaches his voluntary limit for the ligaments, the muscles and the joint stress simultaneously. In other words, the cervical spine structure is best used when all its members are equally solicited. Note that the estimates derived from this assumption are necessarily conservative, because if this assumption are not true, then some of the structures would not be fully solicited at the time of maximum voluntary effort. Therefore by definition, the simulations based on these estimates will result in a conservative estimate of what the cervical spine can support.

The second assumption is that the maximum voluntary effort is 2/3 of the ultimate limit. This is based on a study of the lumbar spine by Gracovetsky and Farfan [17]: it is calculated that a weightlifter will not voluntarily execute a lift that will solicit his lumbar spine components at more than 2/3 of their ultimate strength. Using this information the maximum voluntary muscular effort is taken as 5.8 Kg/cm^2.

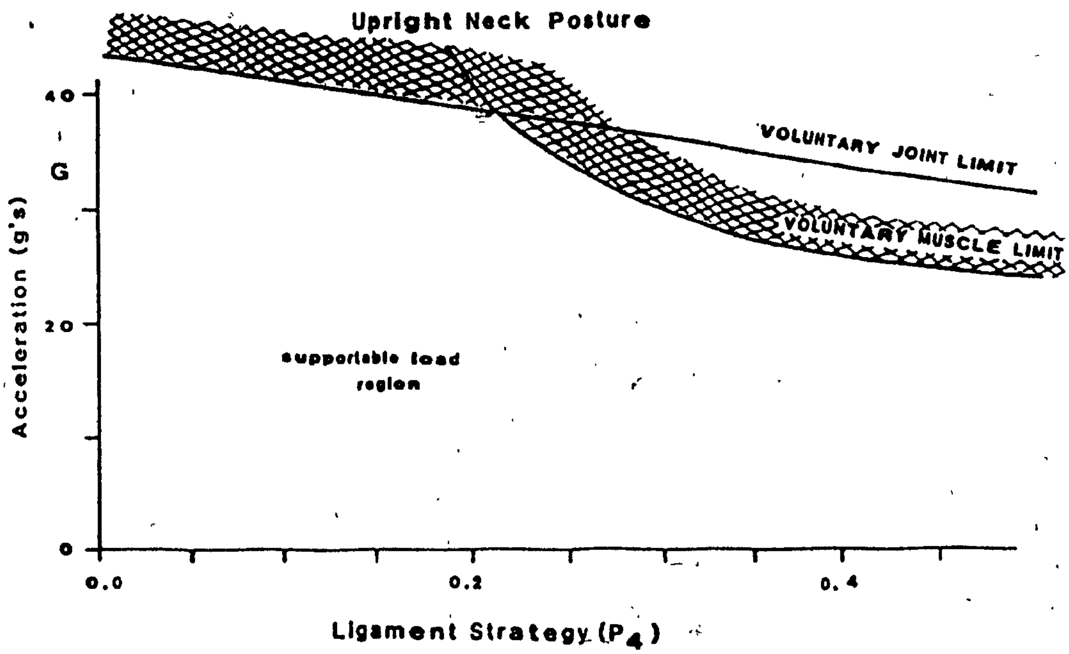
Using the experimental and simulation results of Chapters 2 & 3, the following estimated values of voluntary limit in the normal upright posture can be obtained.

1. joint stress - 160kg
2. ligament moment - 160kg-cm
3. muscle tension - 5.8kg/cm^2

For the fully flexed posture the following voluntary limits can be obtained.

1. joint stress - 160kg
2. ligament moment - 320kg-cm
3. muscle tension - 5.8kg/cm^2

Recall that the objective function has four parameters P_1 to P_4. The parameter which has the greatest effect on the ability of the cervical spine to support an acceleration load is the ligament weighting parameter P_4. The effect of this parameter on the response of the model is best illustrated by a plot of the maximum supportable acceleration (i.e resulting in any spinal member being solicited at 2/3 of its ultimate strength) versus P_4. This has been done for two neck postures and two acceleration directions. (see Fig. 4-4 and Fig 4-5). These figures show the area in which a solution may be found.



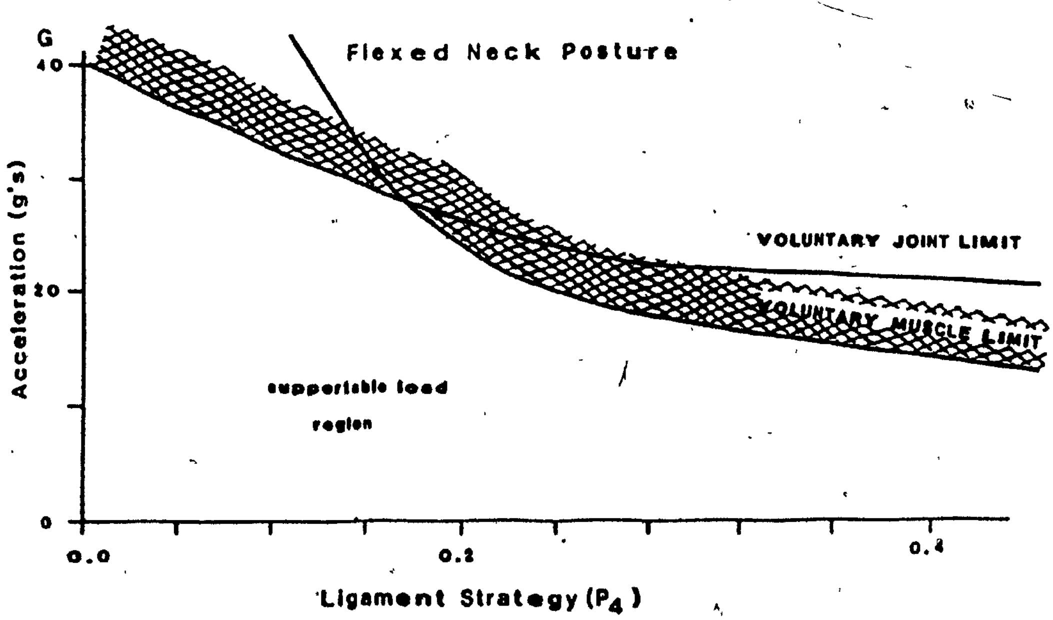
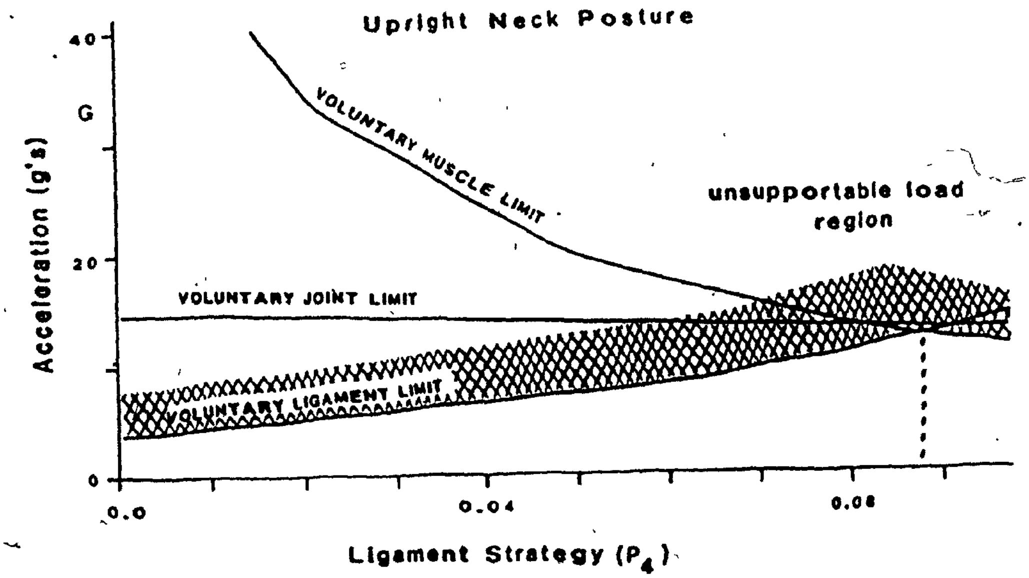


Figure 4-4 Maximum voluntary limits for acceleration loading in the normal upright neck posture and fully flexed neck posture with acceleration load acting approximately through the axis of the spine



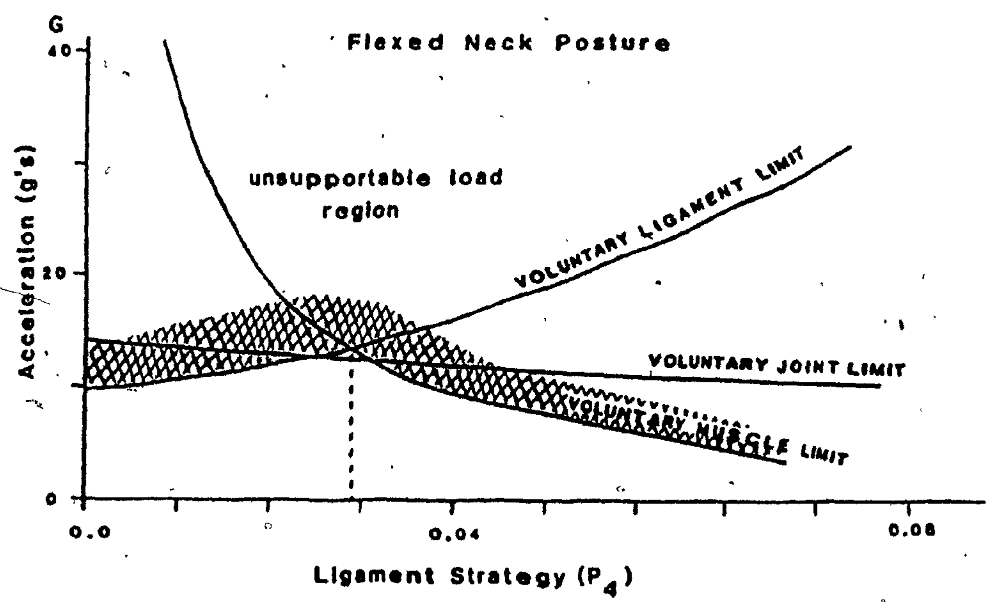


Figure 4-5 Maximum voluntary limits for acceleration loading in the normal upright neck posture and fully flexed neck posture with load acting perpendicular to axis of spine

Normal upright posture - Flexed posture For the cases where the acceleration is acting through the spine, the limiting factor is the joint stress limit. This results in maximum supportable accelerations of 40g and 37g for the normal upright and the fully flexed postures respectively (Fig. 4-4). When the acceleration direction is acting at 90 degrees of the compression direction, a maximum supportable value of 13 g's is calculated for both postures (Fig. 4-5). The values of P_4 are different, namely .0875 and .028 for the upright and flexed postures respectively.

From these results it can be seen that when the acceleration is producing high flexion loads, the voluntary joint limit (maximum joint stress), the voluntary muscle limit and the voluntary ligament limit intersect for, approximately the same value of P_4 that is obtained from the simulation of the experiments of Chapter 3. This result suggests that the extension of the results obtained from the experimental loading arrangement to an acceleration loading arrangement similar to that experienced during pilot ejection is valid.

For the case when the load is acting through the spine, the factor which limits the acceleration which can be supported becomes the voluntary joint limit. This is a result of the fact that the load is almost purely compressive.

The results also indicate that the maximum supportable acceleration does not change appreciably with posture. This is in sharp contrast to the results obtained for muscle strategy in which large changes are observed with change in posture. This result suggests that when the muscles are at their least effective, the ligaments take a larger portion of the load.

4.3 DISCUSSION

The objectives of these simulations are to determine how the head and neck would respond to acceleration loads similar to those experienced in pilot ejection and to obtain estimates of what the maximum acceleration that can be supported. These objectives are pursued in two different manners.

The first is to assume that the neck is being supported primarily by the muscles which are allowed to develop tensions approaching their ultimate limit. Since the experimental results are not used in the assigning of the weighting values (P_1, P_2, etc), it is not felt that the results of these simulations would accurately represent the actual response of the cervical muscles. The passive resistance of the cervical spine would play a much greater role than is the case in these simulations. These results are therefore felt to represent a conservative estimate of the maximum acceleration loads which the neck could support if subjected to the loading configuations of the simulations since the load would be under full muscle control.

There are a number of points about the muscle strategy results which are noteworthy:

1. A high acceleration load can be supported if the load is maintained in such a way that it produces very little flexion or extension moments (i.e. the load acting approximately through the axis of the spine.)

2.The increase in flexion or extension moments (or decrease in maximum supportable acceleration) is very large for only small changes in the direction of loading in the vicinity of the optimum accleration direction. This would suggest the maximum accelerations of 30g, 24g, and 15g for the neutral, flexed and extended postures may not be realisable since the subject may not be able to maintain the desired orientaion with respect to the load.

3. The muscles are most effective in supporting the load in the neutral neck posture. This is a result of the fact that it is possible to obtain lower flexion and extension moments at the intervertebral joint in this position and because the muscles have a slightly better mechanical advantage in this posture.

The second approach for estimating the maximum acceleration load (the ligament strategy) consists of performing simulations using results of the tuned model of Chapter 4. The results of these simulations are a better representation of the the actual response of the neck to acceleration loading than is the case for the muscle strategy simulations since it is based on human experiments. It is felt that these results are still conservative since in the experiments the subjects exerted themselves only to their voluntary limits. Under extreme conditions such as pilot ejection, one may push oneself closer to their ultimate limit.

There are a number of points about the ligament strategy results which are note worthy:

1. When the ligament system is used in conjunction with the muscular system, then the maximum supportable acceleration can be as high as 40 g's in the best case and drops to 13 g's in the worst case. This is a significant improvement over the pure muscular strategy.

2. It is noted that the sharing of the load between the muscles and ligaments is a function of the posture. Nevertheless, the maximum supportable acceleration does not change appreciably. This result may be attributed to the fact that the cervical spine is indeed optimized for any task over its full range of motion. The consistency of this result with the principles on which the study is based is seen as a confirmation that this approach is basically valid.

CHAPTER 5 CONCLUSIONS

The anatomical arrangement of the muscles, ligaments and bones of the cervical spine, including the skull and the shoulder girdle have been described in mathematical terms. The relationship between the external loads and the internal forces generated by muscle and ligament at each intervertebral joint of the spine has been expressed in mathematical terms.

In order to reduce the size of the problem (number of variables), the muscle strands in the cervical spine have been arranged into a number of functional groups, each with its own independent muscular firing density. The muscular strategy required to perform a given task, such as the balancing of an external load, is calculated by minimizing an objective function which is based on the stress levels in the intervertebral joint, the ligaments and the muscles. The objective function expresses the fact that the muscular action balances the external load while attempting to 1) equalize the stress distribution among the intervertebral joints and 2) minimize the stress at the intervertebral joint so as not to exceed the biological limits of material.

The muscular activity can be measured in the volunteers performing isometric tasks and can be used to arrive at values for empirical constants (P_1, P_2, ect) in the model. This process is referred to as tuning.

The model has been tuned primarily to the pattern of EMG response of three of the accessible muscles; the omohyoid, the splenius capitis and the semispinalis capitis. These muscles are suitable because their onset of activity occurs at different levels of extensor effort.

In the process of tuning the model to the experimental results a number of conclusions about the normal functioning of the cervical spine and about methods of analyzing it become apparent:

1) Data can be obtained from electromyographic experimentation which is sufficiently quantitative to allow comparison with the numerical results of a mathematical modelling procedure. Moreover, sufficient information can be acquired by monitoring only superficial muscles with the use of surface electrodes.

2) When the model is utilized to simulate the subjects extending their head against resistance, it is found that the maximum voluntary effort which can be achieved is dictated by the amount of flexion moment that can be generated at the lower cervical joints. This fact is reflected in the model's attempt to recruit the omohyoid muscle when the neck is subjected to very high flexion moments.

3) The ability of the model to successfully simulate the experimental tasks of this study is seen as validation of the basic principles on which the model is based.

4) The preferred direction of loading for the occipital-atlas-axis joint (O/C1/C2) is in the compressive direction. This is indicated by

the fact that the stress at these levels has a very direct and dominant effect on the muscle firing strategy of the cervical spine. This muscle strategy gives rise to a force resultant passing through the O/C1/C2 structure which has a minimal shear component. This observation would seem to suggest that the unrestrained head tends to orient itself into the direction of acceleration.

5) Larger portions of the spine must be considered if an understanding of the role of all the neck muscles is to be obtained. This conclusion is illustrated by the fact that the low level activity observed in the scalene muscles and sternomastoid muscle can not be rationalized in terms of the load which the neck must support.

The drawback in analyzing the spine on a large scale is that the complexity and the large number of assumptions required will make it difficult to draw meaningful conclusions.

The tuned model is then utilized to simulate high acceleration loads using two approaches. First, the case in which the muscles have a dominant role in balancing the acceleration load is analyzed. This approach is considered as a worst case analysis since the possible support which might be provided by the passive resistance of the spine is ignored. The advantage of such an approach is that the tuning of the model is not dependant on experimental results. The disadvantage is that the results may tend to be too conservative.

Second, the case in which the ligamentous structure plays a more dominant role in balancing the load is analyzed . The maximum supportable acceleration is based on the maximum extensor effort achieved by the volunteers in the experimenatal test. For this reason, results obtained here also tend to be conservative. These results are felt to be more realistic than those obtained from the previous approach since the effects of the passive resistance of the cervical spine have been introduced.

Conclusions and suggestions for future work based on the results of the simulations can be summarized as follows:

1) It is clear that the muscle system plays a crucial role in the controlling of the stress level in the cervical spine. This modelling approach is unique in its ability to simulate the response of the muscles. The inability of the discrete parameter models to account for the contribution of the muscles represents a major shortcoming. However, in a situation where the muscles are incapable of supporting the acceleration load, the discrete parameter models may adequately represent the dynamic response of the spine.

2) The maximum supportable acceleration is a function of two elements:

a)- The neck posture assumed during acceleration

b)- The relative alignment of the spinal lordosis to the acceleration vector.

Any ejection analysis which does not consider these two conditions cannot accurately determine the role of the muscular system or the full capability of the musculo-skeletal system in supporting an acceleration load.

3) If it is assumed that the ligament system is not used (eg. the posterior ligament system is relaxed), then the load is entirely balanced by muscular action. For this case, the optimum combination of neck orientation and acceleration vector occurs when the load is acting approximately through the axis of the spine. If it is assumed that the muscles can be recruited to their ultimate limit, a maximum of 30g of acceleration is calculated. If the maximum supportable acceleration is considered to be 2/3 of this ultimate limit, a maximum supportable acceleration of 20g is obtained. Note however that if the spine deviates from the optimum posture, then the maximum supportable acceleration drops to about 5 g.

4) If the passive resistance of the ligaments is considered to be making a significant contribution to balancing the load and all the structural components (muscle, ligament, and joint) are allowed to go to their voluntary limits then the neck can support 40g of acceleration in an optimum loading configuration. For less optimal loading configurations, accelerations in the range of 15g can be supported.

If one assumes that the voluntary limit is 2/3 of the ultimate limit, as was shown in the study of Gracovetsky et al [17], then by extrapolation the estimate of supportable accelerations may be increased from 15g - 40g to 32g - 60 g. However, caution must be exercised before accepting such an extrapolation. For safety reasons, the experimental tuning of the model is performed on human volunteers who had full control of their loading conditions. The actual case for pilot ejection is significantly different from the experimental one used in this study.

5) This method of analysis could be applied to the previous model of Gracovetsky et al on the lumbar spine in order to re-evaluate the ejection problem for the total spine.

6) The principles and methods used in this study could be extended to other problems in which the response of the human spine to acceleration loads is of interest. An example of this is the new generation of fighter aircrafts that subject the pilot to severe lateral forces. This type of analysis could lead to compensatory measures resulting in an improved seat design.

Caution must be exercised however in interpreting the results calculated for high acceleration loads. The magnitude of the loads experienced by the experimental subject is in the same range as that which would be experienced during pilot ejection. However, the experimental load is increased slowly and always under the subjects control. The acceleration experienced by the pilots during ejection is not increased slowly but rather the acceleration reaches high levels in a very short period of time. The response time of the pilot may not be fast enough to allow him to maintain control of his loading conditions. Therefore to determine if it is valid to apply these results to a loading situation such as pilot ejection, appropriate animal experiments and human experiments with safe loading conditions are necessary.

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APPENDIX - NUMERICAL DESCRIPTION OF CERVICAL SPINE

This section describes the mathematical model of the cervical spine which has been used throughout the report. The model describes the geometrical arrangement of vertebrae and skull, the ligaments, and the muscles. It relates the action of the muscles and the load applied to the spine, to the stress at the intervertebral joints.

To avoid the necessity of introducing a large number of assumed parameters, it is desirable to limit the complexity by which the passive spinal structures, such as the joint and the ligaments, are described.

A force balance is imposed on each of the intervertebral joints being considered. The forces acting at a joint can be considered to be due to one of four sources:

1) External Load - force due to the weight of head and neck or an externally applied load.

2) Muscle load - force due to the muscles when contracted.

3) Ligament Reaction - force due to the deformation of the ligament during motion.

4) Joint Reaction - force due to the deformation of the disc or reaction force due to the facet.

Since there are no other forces acting on the joint, the moments, the shear, and compression due to these forces must balance. This can be stated in equation form as follows:

$$\sum_{k=1}^{Nm}(A_{ijk}\ K_k)+E_{ij}+L_{ij}+J_{ij}=0 \qquad (4)$$

A_{ijk}= force component-i (compression, shear, and moment) at joint-j (C2-C3, C3-C4,...,C7-T1) due to a unit stress in muscle group k (multifidus,scalene,etc.)

K_k = stress in muscle group k

E_{ij}= force component i at joint j due to an external load

L_{ij}= component i at joint j due to the reaction of the ligaments at joint j

J_{ij}= force component i of the reaction at joint j due to the disc and the facet

The A_{ijk} are determined from anatomical descriptions of the cervical spine and x-rays; and the values for E_{ij} are obtained from a precise description of the task to be performed.

A.1 DESCRIPTION OF RELEVANT SKELETAL STRUCTURE

The process of numerically describing the muscles consists of determining the points of attachment of the muscles to the relevant skeletal structures. These values are used to determine the resulting reaction forces that these muscles can produce at the I.V. joints.

The relevant skeletal structures consist of the vertebrae of the neck (C1-C7) and the occipital bone and since some of the muscles inserting into the neck and head arise from the thoracic region, the thoracic vertebrae (T1-T6) are also included. Since the motion of the vertebrae in the thoracic region is very small compared to those in the cervical region, the thoracic vertebrae are assumed to be a part of a fixed base on which the neck operates.

In addition to the thoracic vertebrae, muscles of the neck are also arise from the ribs, clavicle, sternum, and scapula. The sternum and ribs can be thought to be a part of a fixed base. The scapula--clavicle structure is moveable and can change the line of action of the muscles arising from it. For this study the shoulders are assumed in a fixed position and the arms unrestrained.

Each of the vertebrae, ribs, and bones is treated as a rigid body. The points of interest on the rigid body structures are described in a local coordinate frame and the skeletal structures are assembled by assigning a location and orientation to each of the local coordinate frames.

The geometric information is collected from a number of sources. These include x-rays of an individual in three postures (upright or neutral, full flexion, and full extension), an assembled skeleton, and individual vertebrae. The points of attachment to the vertebrae are organized in a single array [3 x 17]. Each of the points is assigned a code number as follows:

code number = m x 100 + n

where m = rigid body level

n = point number for that rigid body

The location and description of these points are presented in the remainder of this section.

OCCIPITAL BONE - For the occipital bone, it is found to be convenient to locate the origin of its local coordinate frame at the centre of the external auditory meatus and define the y-axis as running in the direction of the line from the floor of the orbit to the external auditory meatus. The z-axis is taken as running up through the top of the skull. The bone is described by 18 points in this local coordinate frame. These points represent the mean points of insertion of the muscles which attach to the head. See Table A-1.

ATLAS (C1) - The atlas is described by a coordinate frame which has its x-axis running from the anterior tubercle to the tip of the spinous process and the z-axis running upwards from the anterior tubercle. This vertebra is described by four points in its local coordinate frame. See Table A-2.

AXIS (C2) - The origin of the local coordinate frame for the Axis is taken at the inferior anterior corner of the body with the z-axis running up to the anterior articular surface of the dens. The y-axis runs posteriorly from the origin. The Axis is described by six points.

Point No.		Point Description
101	-	insertion: rectus capitis post. major
102	-	insertion: rectus capitis post. minor
103	-	insertion: oblique capitis superior
104	-	insertion: rectus capitis lateral
105	-	insertion : rectus capitis anterior
106	-	insertion: longus capitis
107	-	insertion: sterno-mastoid
108	-	insertion: trapezius
109	-	insertion: longissimus capitis
110	-	insertion: semispinalis capitis
111	-	insertion: splenius capitis from C4
112	-	insertion: splenius capitis from C5
113	-	insertion: splenius capitis from C6
114	-	insertion: splenius capitis from C7
115	-	insertion: splenius capitis from T1
116	-	insertion: splenius capitis from T2
117	-	insertion: splenius capitis from T3
118	-	insertion: splenius capitis from T4

Table A-1 Points of attachment of muscles to the occipital bone (Points of attachment shown in Fig. A-1)

Point No.		Point Description
201	-	insertion: longus colli
202	-	origin: rectus capitis lateral
		rectus capitis anterior
203	-	insertion: oblique capitis inferior
		origin: oblique capitis superior
204	-	origin: rectus capitis posterior
minor		

ATLAS (C1) (see Fig. A-1)

Point No.		Point Description
301	-	anterior inferior corner of vertebra body
302	-	posterior inferior corner of vertebra body
303	-	insertion: longus colli
304	-	anterior tubercle
305	-	posterior tubercle
306	-	tip of spinous process

Axis (C2) (see Fig. A-2)

Table A-2 Points of attachment of muscles to the atlas and axis

Point No.		Point Description
m01	-	anterior inferior corner of vertebra body
m02	-	posterior inferior corner of vertebra body
m03	-	posterior superior corner of vertebra body
m04	-	anterior superior corner of vertebra body
m05	-	anterior surface of vertebra body
		origin & insertion: longus colli
m06	-	anterior tubercle
m07	-	anterior tubercle
m08	-	articular process
m09	-	anterior portion of spinous process
m10	-	middle portion of spinous process
m11	-	tip of spinous process

m = vertebra level (i.e. m=4 to 8 corresponds to C3 to C7)

Table A-3 Points of attachment of muscles to cervical vertebra (C3 - C7) (see Fig. A-2)

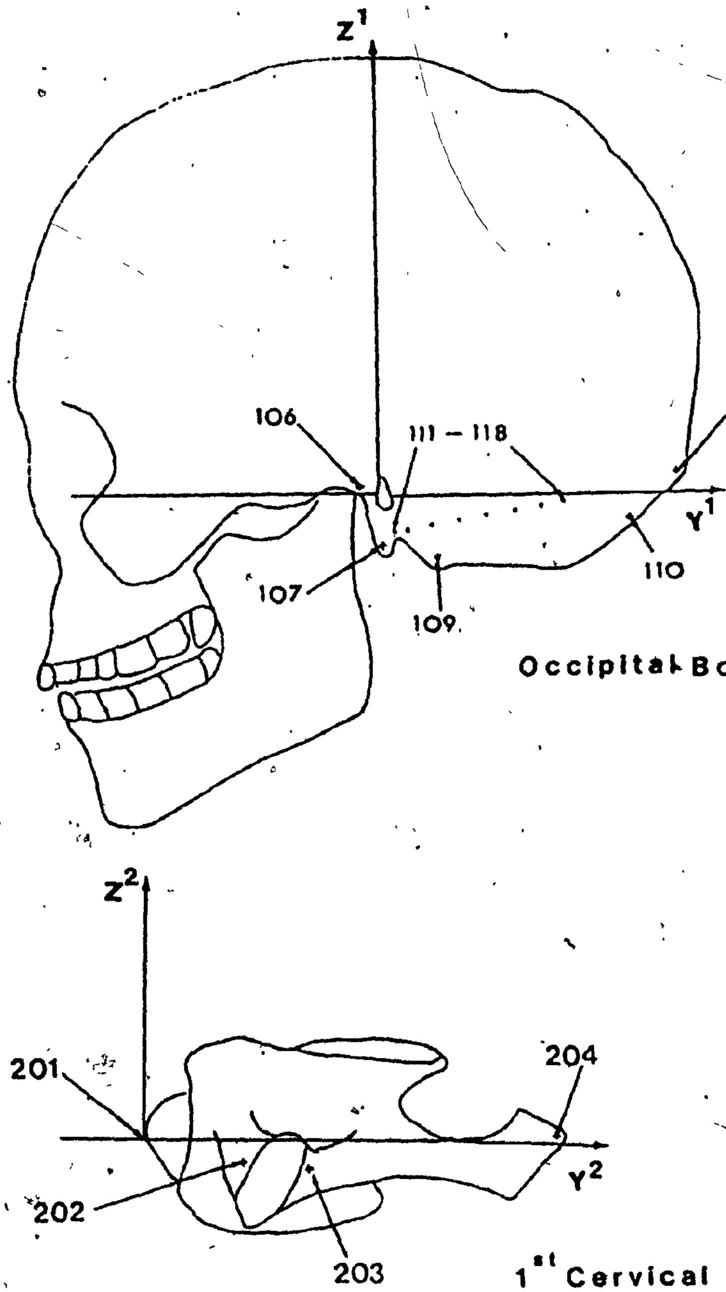


Figure A-1 Points of muscle attachment on the head and atlas

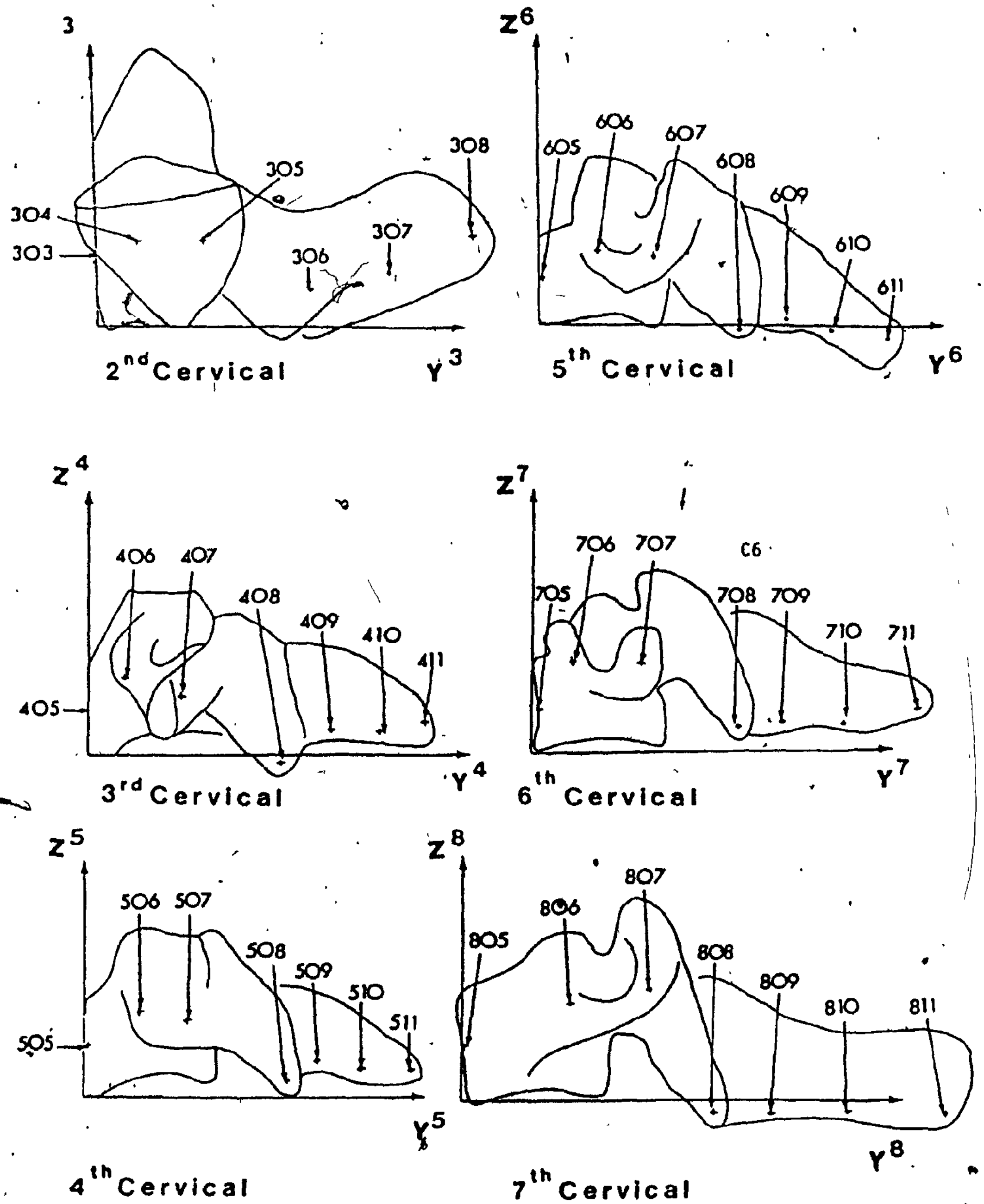


Figure A-2 Points of muscle attachment on the 2nd to 7th cervical vertebra

CERVICAL VERTEBRA (C3-C7) - The origin is located at the inferior anterior corner of the body with the z-axis running up to the superior anterior corner. The y-axis runs posteriorly from the origin. All of these rigid bodies are described with 11 points in the local coordinate frame. See Table A-4.

THORACIC VERTEBRA (T1-T6) - The rest of the relevant skeletal structures may be assumed to be a part of a fixed base and can therefore be described in terms of a single coordinate frame. It is felt, however, that it is be advantageous to assign a local coordinate frame to each of the thoracic vertebrae.

The thoracic vertebrae can not be viewed in lateral X-ray because their view is obstructed by the arms and ribs. The points of interest on each of the vertebra are determined from skeletal specimens. Each vertebra is described in its own local coordinate frame (Table A-5).

The local coordinate frames for the thoracic vertebrae are defined in the same manner as for the cervical vertebrae. These vertebrae are described with eleven points.

Sternum - Clavicle - Scapula - The sternum is described as a single point which is used for the origin of the local coordinate frame of the fixed base and is also used as the origin of the global coordinate frame. The clavicle and scapula are assumed to be fixed in a known position and all points of muscle attachment are described in terms of the coordinate frame located at the sternum (Table A-6).

Point No.		Point Description
M01	-	anterior inferior corner of vertebra body
M02	-	posterior inferior corner of vertebra body
M03	-	posterior superior corner of vertebra body
m04	-	anterior superior corner of vertebra body
m05	-	anterior surface of vertebra body
m06	-	middle portion of transverse process
m07	-	base of transverse process
m08	-	tip of transverse process
m09	-	middle portion of spinous process
m10	-	tip of spinous process
m11	-	angle of rib

m = vertebra level (i.e. m=9 to 14 corresponds to T1 to T6)

Table A-4 Points of attachment of muscles to thoracic vertebra (T1 - T6) (see Fig. A-3 and A-4)

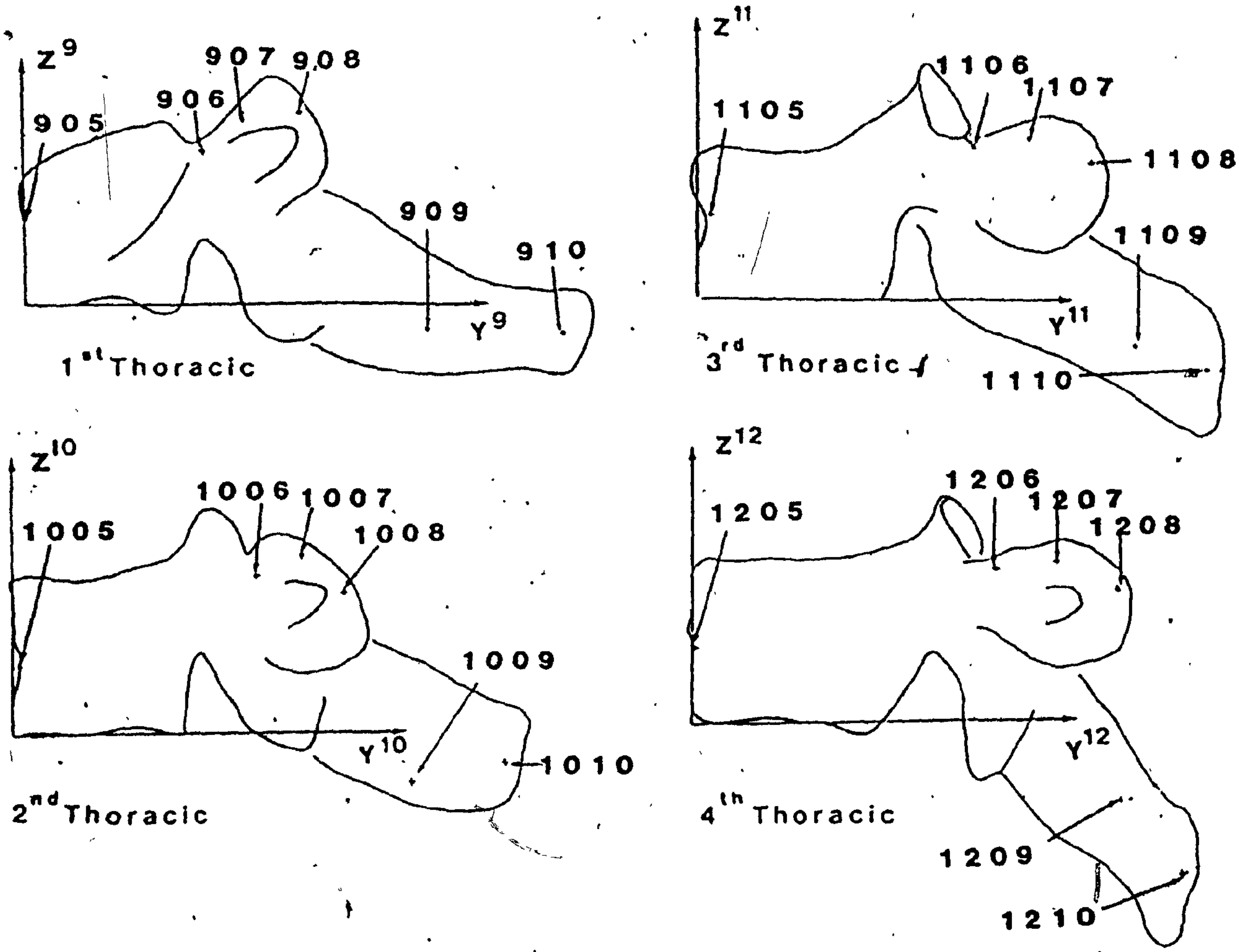


Figure A-3 Points of muscle attachment for the 1st through 4th thoracic vertebra

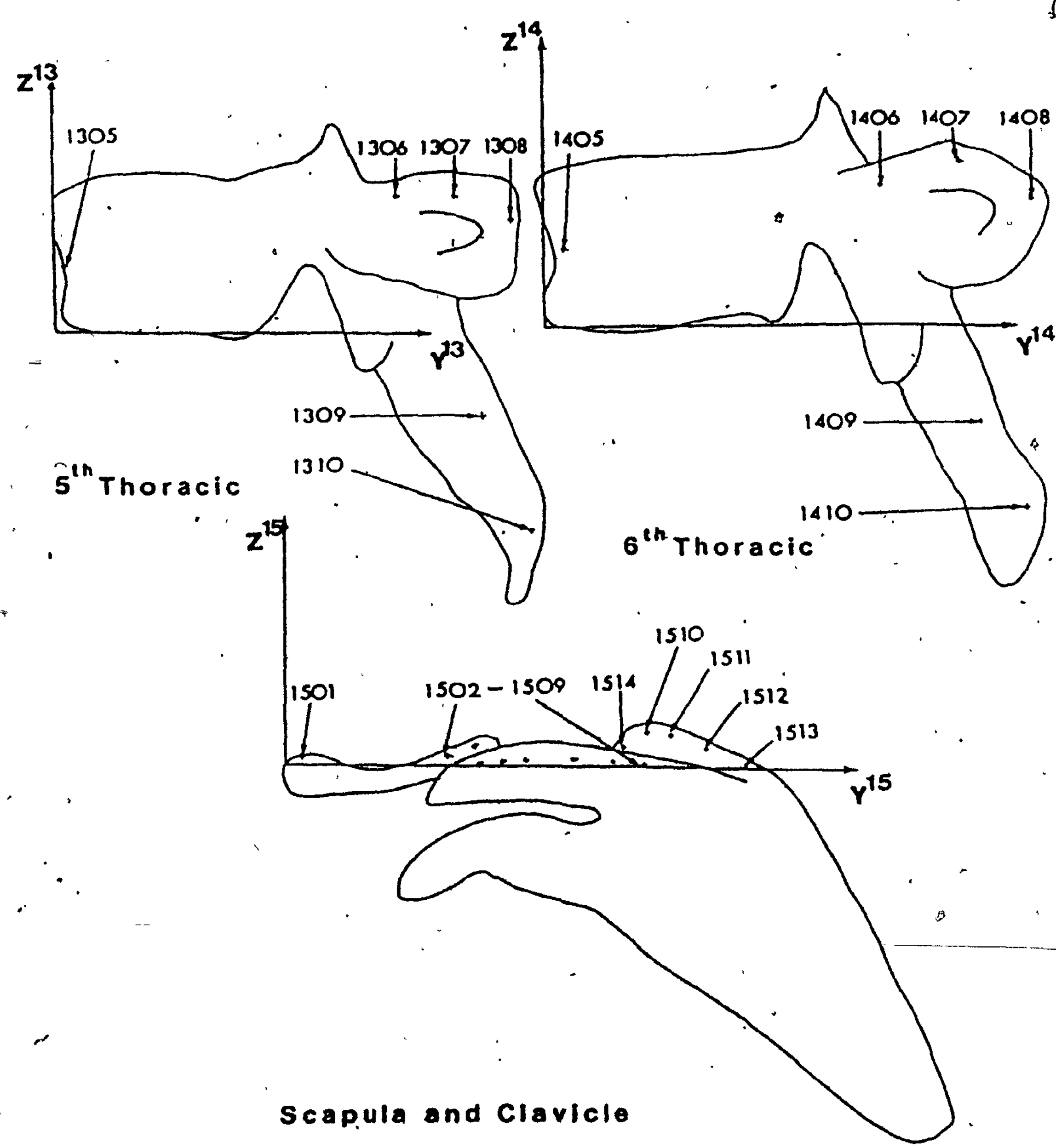


Figure A-4 Points of muscle attachment for the 5th and 6th thoracic vertebra and scapula and clavicle

Point no.		Point Description
1501	-	anterior end of clavicle
1502	-	posterior end of clavicle origin : trapezius to occipital bone
1503	-	posterior end of clavicle origin: trapezius to CL
1504	-	acromion, origin: trapezius to C2
1505	-	acromion, origin: trapezius to C3
1506	-	acromion, origin: trapezius to C4
1507	-	scapula spine, origin: trapezius to C5
1508	-	scapula spine, origin: trapezius to C6
1509	-	scapula spine, origin: trapezius to C7
1510	-	vertebral margin of scapula origin: levator-scapula to C3-C4
1511	-	vertebral margin of scapula origin levator-scapula to C2-C1
1512	-	vertebral margin of scapula origin: rhomboid to C7
1513	-	vertebral margin of scapula origin: rhomboid to C6
1514	-	origin: omohyoid

Table A-5 Points of attachment of muscles to clavicle and scapula (see Fig. A-4)

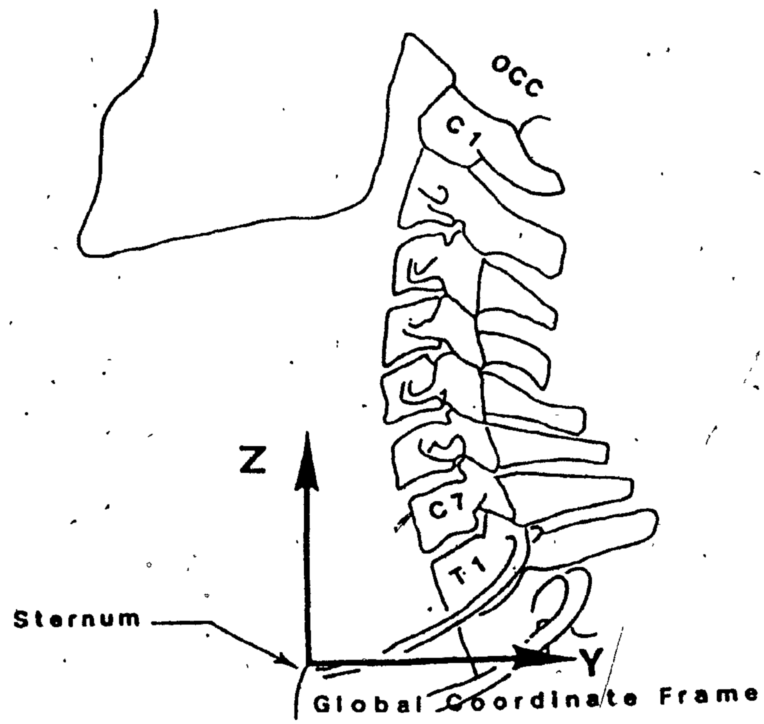
Ribs - Only the first three ribs are considered in this structure since all muscles (with the exception of iliocostalis) traversing the neck originate above the third rib. Iliocostalis, however, is described in the thoracic coordinate frame.

Transformation to Global Coordinate Frame - Having assigned a set of coordinates to each of the points of interest in their repective local coordinate frames, it is now possible to transform them to a global frame. The spinal geometry is different for all neck postures, therefore three representive postures are choosen (neutral, flexion, and extension). Using the location of the origin and the orientation of each of the local coordinate frames, the local coordinates are transformed to a global frame located at the sternum (Fig. A-5).

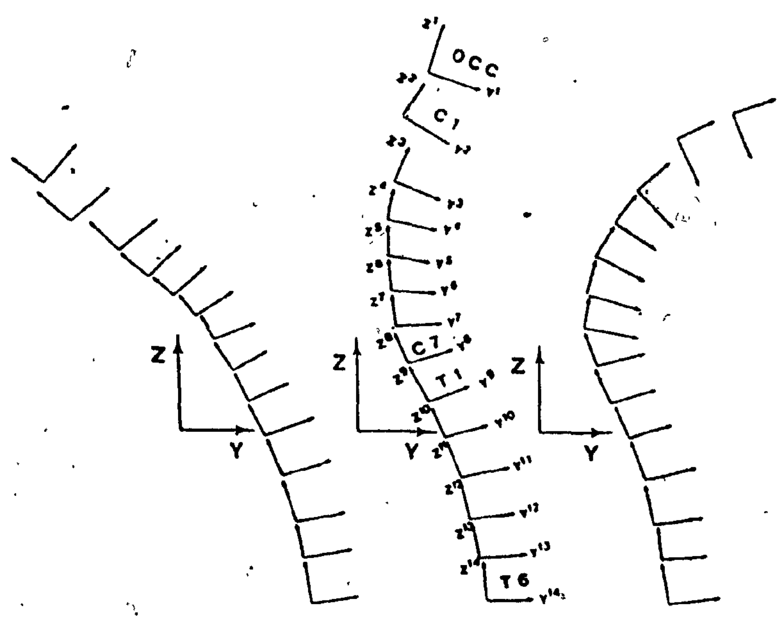
A.2 LIGAMENT DESCRIPTION

The neck is be assumed to be in a flexed ,extended, or normal upright (neutral) posture with the jaw closed at all times. The applied load is be limited to forces acting in a direction (from posterior to anterior) so as to produce flexion moments at the intervertebral joints. No load is applied to the arms and shoulders. The arms are assumed to be adequately supported and therefore do not represent a load to the head and neck structure.

When a load is supported by the spine, a balance must be achieved at each of the intervertebral joints with the use of the active structural elements (ie. muscles), and the passive structural elements (ie. ligaments, intervertebral disc and articular facets).



a) global coordinate frame located at the sternum



b) local coordinate frames in global coordinate frame

Figure A-5 Representation of three basic postures in global frame achieved through coordinate transformation

When a flexion moment acts on the intervertebral joint, it is assumed that the primary compressive structure is the intervertebral disc and the primary passive tensile element is the posterior ligament system. The center of the compressive forces is assumed to be acting at the center of the disc's nucleus. The center of the passive tension is assumed to be acting at the center of area of the posterior ligaments.

When the intervertebral joint undergoes bending, resistance to the motion results from the elongation of the ligament fibers and the deformation of the intervertebral disc. In order for the disc to produce a bending moment it is necessary for the posterior portion of the disc to undergo extension deformation or a lesser degree of compression than the anterior portion of the disc. This concept is illustrated by idealization of the uniform disc and ligament properties (Fig. A-6). This idealization would result in stress concentrations at the anterior portion of the intervertebral disc and the posterior fibers of the ligaments.

The disc is described as a purely compressive element with a uniform distribution of stress, and with the center of compression acting at the center of the disc's nucleus. This description does not allow for a bending moment to be absorbed by the disc. A direct consequence of idealizing the disc in this manner is that the ligaments are the only elements producing a passive resistance to bending deformation. As done with the disc, the stress in the posterior ligament structure is assumed to be uniformly distributed.

The disc and the ligaments are represented as single vectors. In

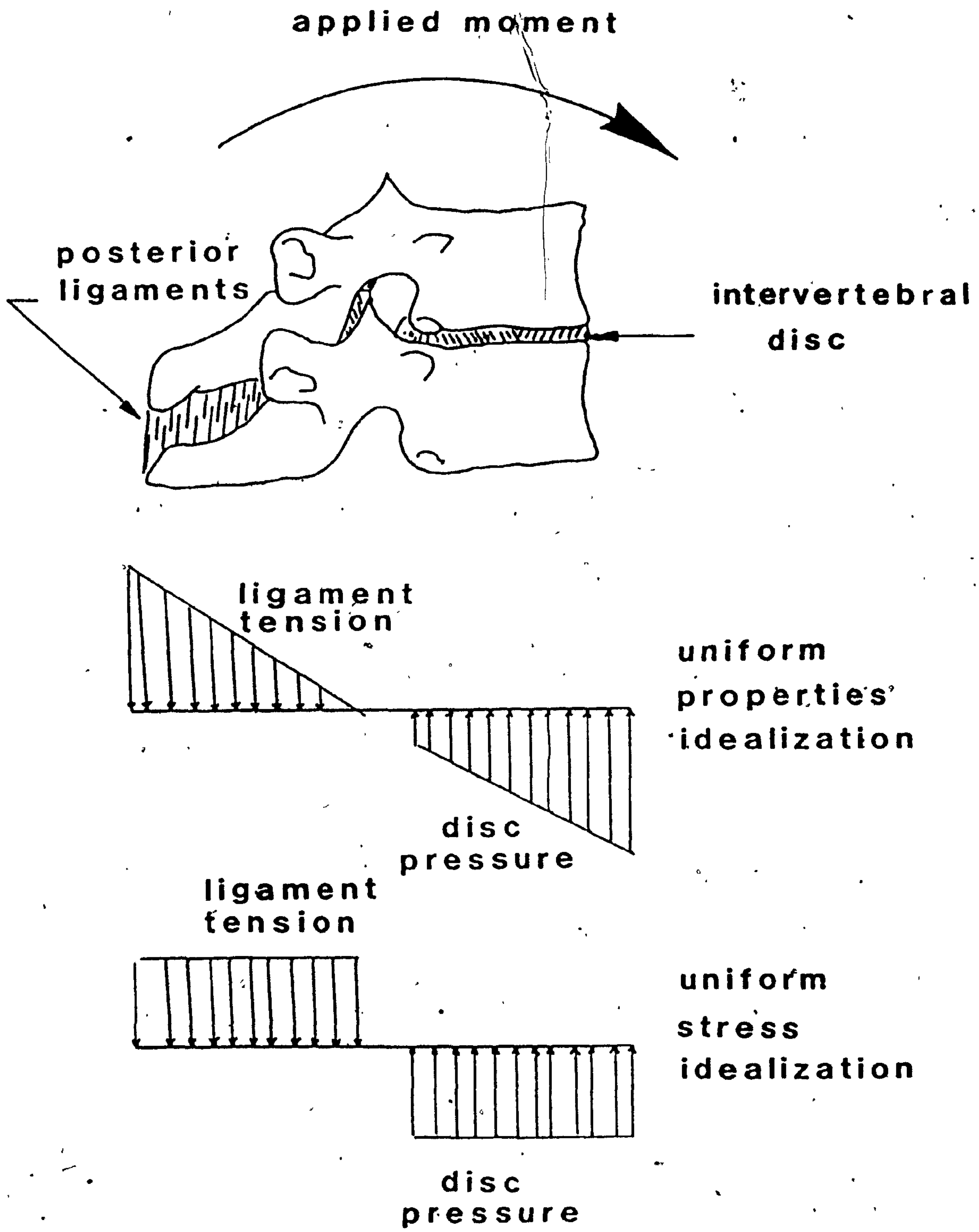


Figure A-6 Idealization of intervertebral joint and ligaments the case of the disc, the vector is perpendicular to the plane which bisects it, and runs through the geometric center. The ligament vector is perpendicular to the disc vector and runs through the center of cross-sectional area. This description results in the ligaments producing only compressive forces in the joint and no shear. It is not possible to determine the exact line of action of the ligament as is the case for the muscles, but examination of the anatomy suggests that the above idealization results in a good approximation.

The ligament's response is a function of the amount of deformation they have undergone. The intuitive approach to modelling this type of structure is to introduce an effective modulus of elasticity or an effective rotational stiffness for each joint. Such an approach would rely on the accuracy of the estimation of this parameter and also on our ability to measure or estimate the degree of bending that the individual joints have undergone. Since one of the main objectives of this modelling approach is to avoid the introduction of these types of assumed or estimated parameters, a different approach based on the use of electromyographic information is proposed.

While standing erect or while fully flexed, the spine requires very little or no muscular activity to maintain posture. This suggests that in these postural positions, the passive resistance of the ligaments is sufficient to support the spine and associated body weight. Using this observation, the ligament tension in the fully flexed posture can be estimate as being that value which balances the load resulting from the weight of the head and neck.

As the load applied to the neck increases, it is reasonable to assume that a portion of this increased load is be taken up by the ligaments. In order for the ligaments to take on an increased load they must undergo some deformation. If the subject maintains an upright neck position as the load increases, the intervertebral joints begin to deform in order to generate the required ligament extension. If the subject attempts to maintain a fixed posture, it is conceivable that the lower joints might undergo a small amount of forward flexion which could then be offset by extension of some of the upper joints.

As a result, the overall posture of the neck remains the same with a small amount of deformation in the spine. Obviously, as the load increases, it becomes difficult to deform the spine and still maintain the same effective posture.

A.3 Description of Intervertebral Joint

The joint is considered as a point located at the center of the disc and is defined as the geometric mean of the anterior and posterior lower corners of the body above the disc and by the anterior and posterior upper corners of the body below the disc. A unit shearing load may be defined as a unit vector acting in the line of the bisector of the disc and is positive in the posterior direction. A unit compression load is defined as being perpendicular to the shear direction, and is positive when acting downwards.

A.4 Muscle Description

The muscles are considered as a collection of muscle strands which are represented by vectors. Each vector runs from a point of origin to a point of insertion. The magnitude of the force produced by each of these idealized muscle strands is equal to the stress developed in the muscle fibers when contracting, multiplied by the cross-sectional area of each of the strands. Therefore for each vector, the shear (A_{1jk}), the compression (A_{2jk}) and the moment (A_{3jk}) produced at each joint is determined. If one considers each strand of a muscle or group of muscles to act at the same stress level when being fired, then the total muscle action can be considered as the sum of the compressions, shears, and moments produced by this group of strands. This may be expressed as follows (Fig. A-7):

$$A_{1jk} = [\underline{S}_j \cdot \underline{M}_k] M_k \qquad \text{(A-1)}$$

$$A_{2jk} = [\underline{C}_j \cdot \underline{M}_k] M_k \qquad \text{(A-2)}$$

$$A_{3jk} = M_k \cdot \mathrm{mag}[\underline{r} \times \underline{M}_k] \qquad \text{(A-3)}$$

where $\underline{r}$ = a vector running perpendicular to muscle strand k and through joint j

$\underline{M}_k$ = a unit vector running from the point of origin of the muscle to its point of insertion

M_k = the cross sectional area of muscle strand k

x = cross product

D_j= geometric center of disc at level j

$\underline{S}_j$= unit vector in shear direction of disc level j

$\underline{C}_j$= unit vector in compression direction of disc level j

$\underline{M}_k$= unit vector in direction of force of muscle strand k

M_k=cross-sectional area of muscle

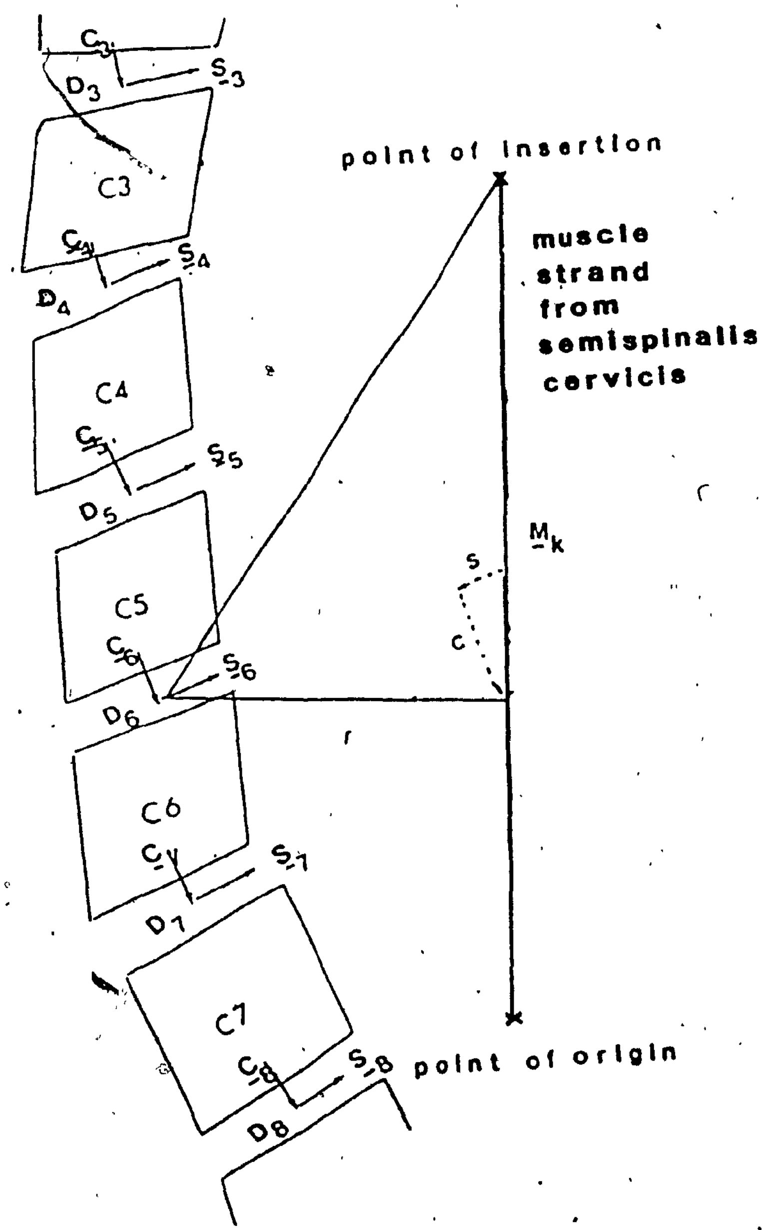


Figure A-7 Muscle and ligament tensions result in shear, compression and moment at the intervertebral joint

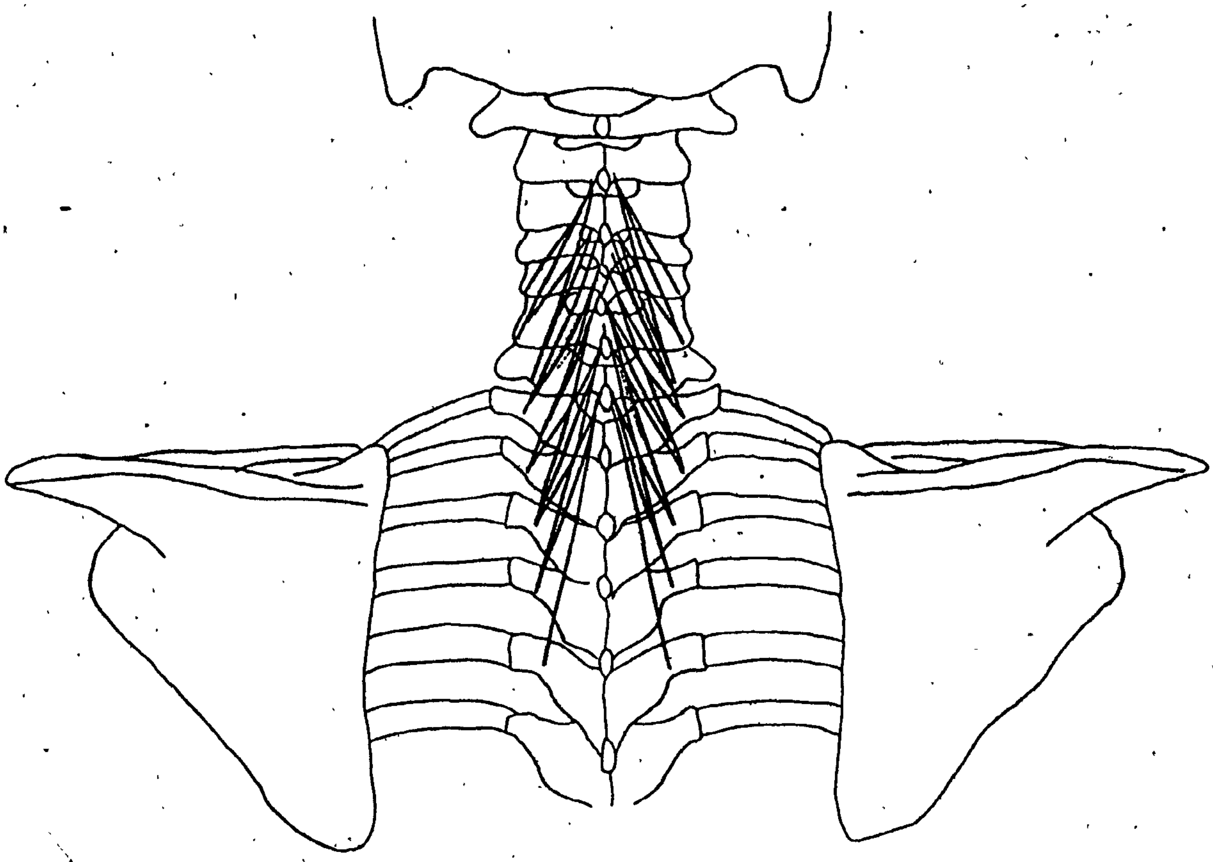
The location of all the points of attachment of the muscles and the location of the local coordinate frame relative to the global frame may be determined from anatomical descriptions, X-rays, and skeletal specimens. With this data the location of the joints and muscle vectors may be determined and hence the muscle matrix (A_{ijk}).

Multifidus - The multifidus is the deepest of the muscles to be considered in this description of the cervical spine. It is noted that the rotatores have a small cross-sectional area and therefore their contribution to balancing the external forces is assumed to be insignificant. The same argument is applied to the inter-transversarius muscles and the inter-spinalis muscles.

The multifidus is a branching muscle having up to 3 penate muscle strands reaching upwards 3,4, and 5 vertebrae above the level of origin. Each of the strands is modelled as a separate vector originating from the lower edge of the articular process and inserting into the spinous process, at 1/3 of the distance from the articular process to the tip of the spinous process (Fig. A-8).

Only those strands which insert into the cervical region are considered. Hence the lowest vertebra of origin for this muscle is T5 and the highest vertebra of insertion is the Axis. The cross-sectional area is obtained at several levels from cross-sectional views of the neck. By assuming equal areas for each of the strands, an area of $0.4cm^2$ is obtained.

This description of the multifidus muscle is applied to the numerical description of the skeleton. Values of length for each of the strands in the neutral, fully flexed, and fully extended positions



Muscle No. 1 Strand No.	Multifidus ID. No.	Origin	Insertion	Area(cm^2)
1	101	608	306	0.4
2	102	708	409	0.4
3	103	708	306	0.4
4	104	808	509	0.4
5	105	808	409	0.4
6	106	808	306	0.4
7	107	908	609	0.4
8	108	908	509	0.4
9	109	908	409	0.4
10	110	1008	709	0.4
11	111	1008	609	0.4
12	112	1008	509	0.4
13	113	1108	809	0.4
14	114	1108	709	0.4
15	115	1108	609	0.4
16	116	1208	809	0.4
17	117	1208	709	0.4
18	118	1308	709	0.4

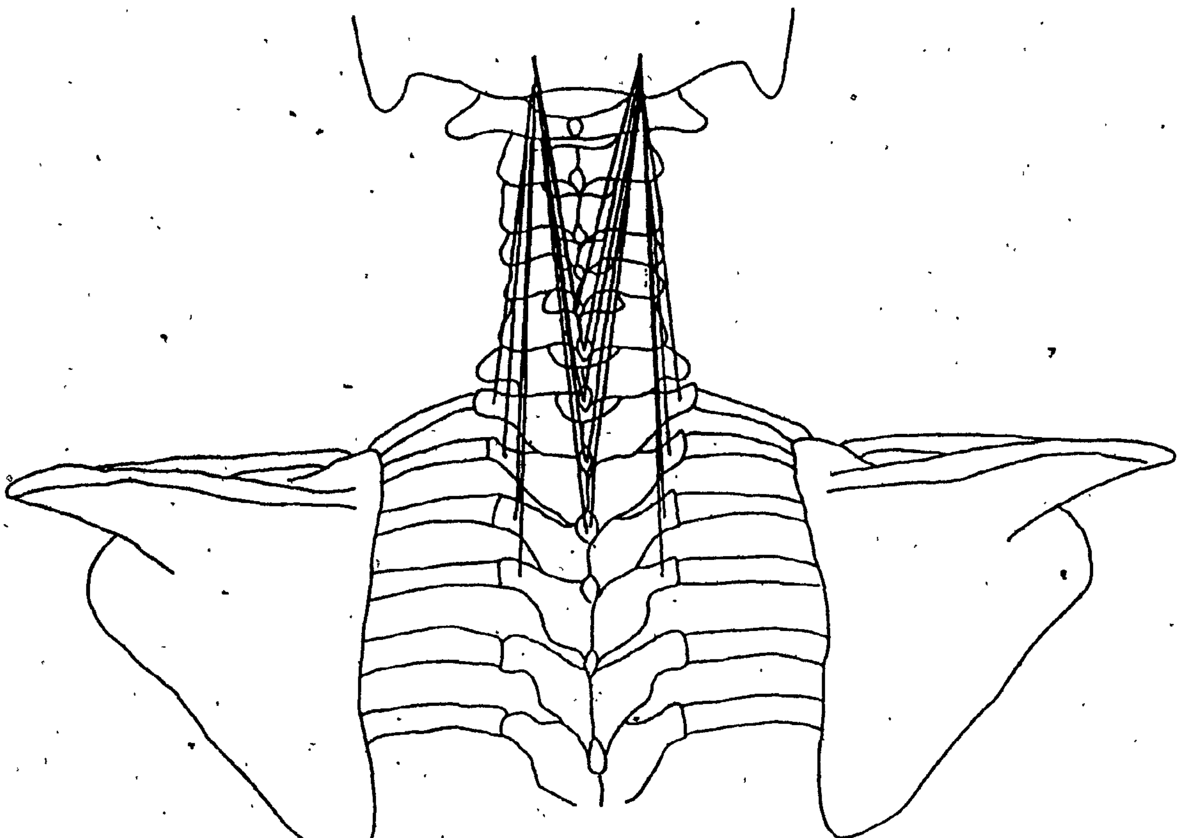
Fig. A-8 Vector description of the multifidus muscle are obtained along with the corresponding increase in length due to flexion and extension. It is known that the maximum change in length of a penate muscle is about 30%. The maximum increase in length due to flexion, and the maximum decrease in length due to extension, as calculated from this numerical description is below 30%. This supports the validity of this description.

The combined effect of all the multifidus strands can be evaluated. It can be shown that the maximum output is produced at the C7-T1 joint and the minimum at the C2-C3 joint. This obvious result can be attributed to the fact that more strands traverse the lower joints than the upper joints.

Semispinalis Capitis and Spinalis Capitis - These two muscles are the largest muscles of the neck. Some anatomy books do not differentiate between these two muscles and treat them as a single mass. These muscles cover the multifidus as it arises from the transverse process of T4 to T1 and the articular processes of C7 to C4 and from the spinous processes of T4-C4 and pass almost straight up the neck to insert between the superior and inferior nuchae lines of the occipital bone.

The portion of this muscle mass arising from the transverse processes of the thoracic vertebrae and articular processes of the cervical vertebrae is termed the semispinalis capitis, and the portion arising from the spinous processes is termed the spinalis capitis.

The semispinalis capitis and the spinalis capitis are modelled as 18 strands arising from the 18 points of origin mentioned above and inserting at a common point of insertion located halfway between the



Muscle No. 2 Semisinalis Capitis

Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
19	201	508	110	0.5
20	202	608	110	0.5
21	203	708	110	0.5
22	204	808	110	0.5
23	205	906	110	0.5
24	206	1006	110	0.5
25	207	1106	110	0.5
26	208	1206	110	0.5
27	209	1306	110	0.5
28	210	510	110	0.5
29	211	610	110	0.5
30	212	710	110	0.5
31	213	810	110	0.5
32	214	909	110	0.5
33	215	1009	110	0.5
34	216	1109	110	0.5
35	217	1209	110	0.5
36	218	1309	110	0.5

Figure A-9 Vector description of semispinalis and spinalis capitis muscles superior and inferior nuchae lines, to the right and left of the center of the occipital bone (Fig A-9).

As with the multifidus, the cross-sectional area per strand is considered equal for all the strands. The above muscle description is appplied to the numerical description of the spine to obtain values of muscle strand length; variation of length due to flexion and extension; and the moment, shear and compression due to a unit stress in each of the strands.

The semispinalis muscle fibers are parallel, as opposed to the penate fibers of the multifidus. It is known that parallel muscle fibres can extend or contract up to 35% of their rest length. The change in length of some of the semispinalis strands are to some extent a little higher than would normally be anticipated, but it is felt that these values are reasonable enough to justify the present description of the muscle.

The large cross-sectional area of these muscles makes them primary candidates for supporting of the neck. The largest moment is produced at the higher joints because of the high cross-sectional area of these muscles at these levels.

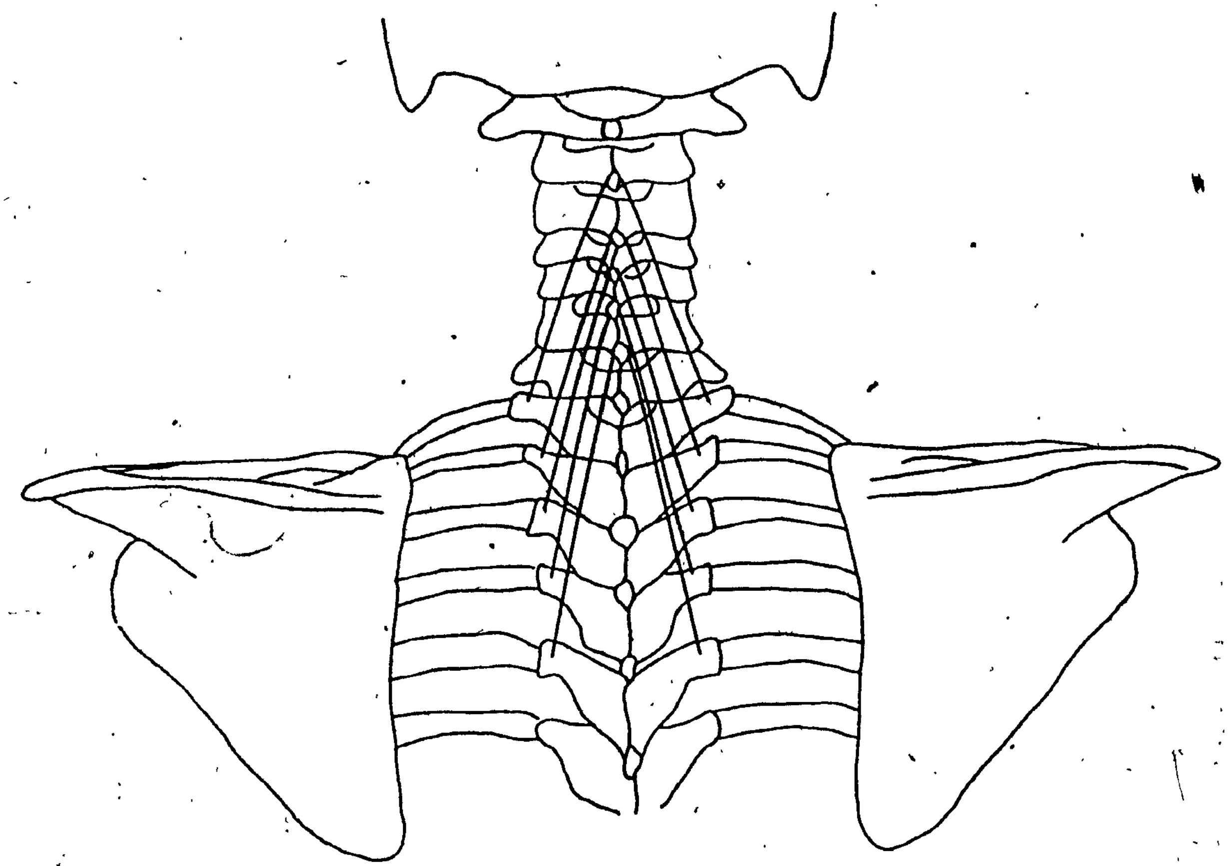
The strands arising from the lower thoracic vertebrae T2-T4 show very little change in length. This is also the case for the multifidus muscle. This can be partly attributed to the fact that the thoracic vertebrae are considered as fixed when in fact they rotate a few degrees in flexion. Another factor is that the thoracic vertebrae are not clearly visible in the lateral x-ray and their position had to be estimated rather than measured. The small change in length may be due to this error. It can also be noted that the small change in length may be attributed to the fact that a greater portion of the muscle strand is tendonous.

Semispinalis Cervicis - The semispinalis cervicis, like the semispinalis capitis, is a parallel fibre muscle but unlike the semispinalis capitis, it has many points of origin and insertion. It arises from the transverse processes of T1-T6 and inserts into the spinous processes of C2-C7 with the longest fibres passing over 6 vertebrae and the shortest passing over 4. For the sake of convenience it is assumed that all of the strands pass over 5 vertebrae. Therefore the semispinalis cervicis may be modelled as 6 strands of equal cross--sectional area with the lowest strand arising from T6 and inserting into C7 and the highest strand arising from T1 and inserting into C2 (Fig. A-10).

The result of this vector description shows that the changes in length of the strands are well within the 35% change that one would expect for parallel fiber muscles. The muscle produces a slightly higher moment to compression ratio than the multifidus. This is due to the fact the muscle lies posteriorly to the multifidus.

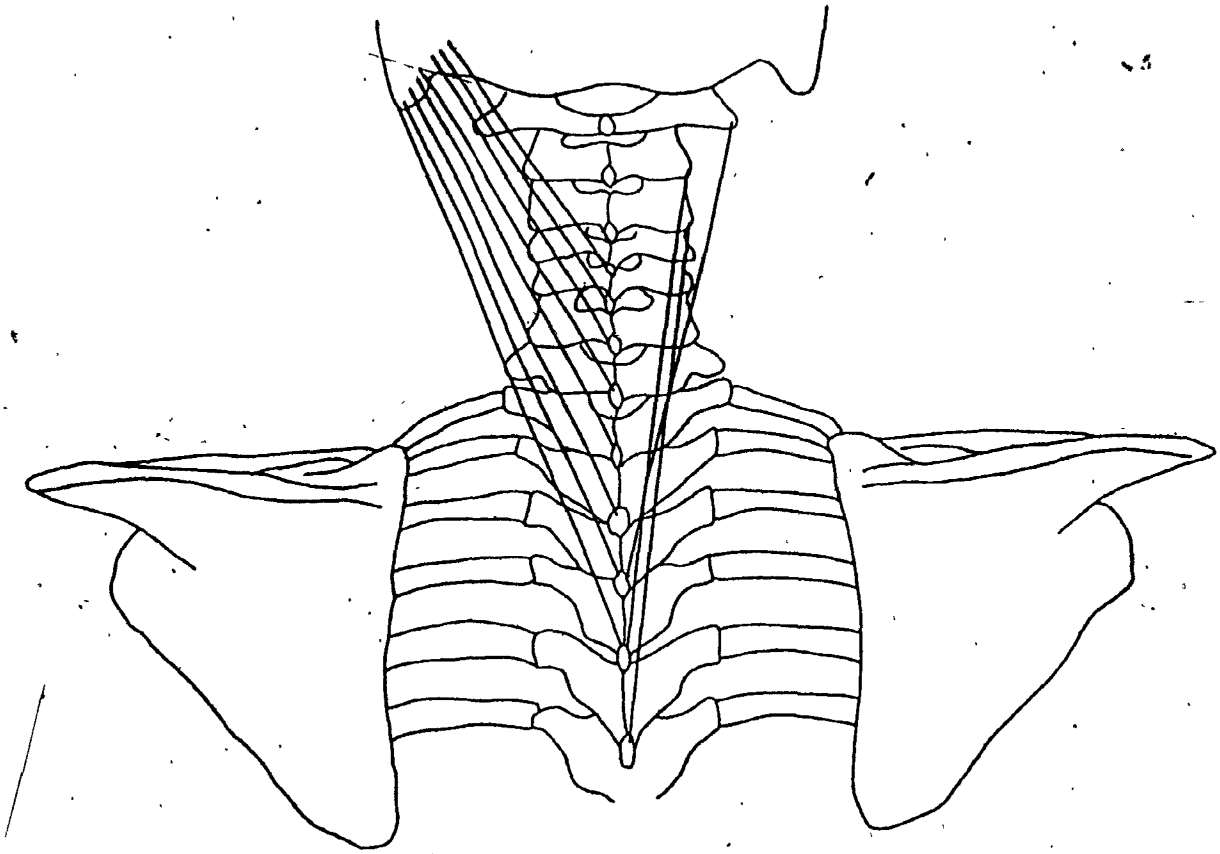
Splenius Capitis - Cervicis - These muscles are the only ones in the neck which originate from the midline and run laterally and upwards. It is presumed that these muscles act primarily in producing torque or act as a torsional stabilizer. (Fig. A-11).

The splenius capitis arises from the ligament nuchae at the level of C4 to C7 and from the spinous processes of T1 to T4; and it



Muscle No. 3	Semisinalis Cervicis			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
37	301	906	310	0.4
38	302	1006	410	0.4
39	303	1106	510	0.4
40	304	1206	610	0.4
41	305	1306	710	0.4
42	306	1406	810	0.4

Figure A-10 Vector description of the semispinalis cervicis muscle



Muscle No. 4	Splenius Capitis			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
43	401	511	118	0.6
44	402	611	117	0.6
45	403	711	116	0.6
46	404	811	115	0.6
47	405	910	114	0.6
48	406	1010	113	0.6
49	407	1110	112	0.6
50	408	1210	111	0.6

Muscle No. 5	Splenius Cervicis			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
51	501	1110	203	0.4
52	501	1110	305	0.4
53	501	1110	407	0.4
54	501	1110	507	0.4

Figure A-11 Vector description of the splenius cervicis and capitis inserts into the occipital bone just below the lateral part of the superior nuchae line and into the mastoid process. The muscle is described in the model with 8 strands. The 4 strands arising from the ligament nuchae are assumed to originate from the tips of the spinous process of C4-C7 and the other 4 strands originate from the tips of the spinous processes of T1 to T4. All 8 strands insert into the occipital bone at 8 individual points of insertion. The strand with the lowest point of origin inserts the most laterally and the uppermost strand inserting more medially.

The splenius cervicis arises from the spinous processes of T3-T6 and inserts into the posterior tubercle of the transverse processes of C1-C4. The muscle is described by 4 strands with the lowest strand arising from T6 and inserting into C4 and the uppermost strand arising from T3 and inserting into C1.

The resulting changes in length for both muscles are well below the 35% value that would be expected for a parallel fibre muscle. Another result worth noting is the large decrease in moment produced when the posture switches from neutral to flexed. This suggests that the role of these muscles may be that of torsional stabilizers rather than forward moment supporters.

Longissimus Capitis - Cervicis - The longissimus capitis is a slender muscle originating from a)the articular processes of C3 to C7 and b)from the tips of the transverse processes at T1 to T3, and inserting into the occipital bone at the posterior margin of the mastoid process just under the splenius capitis. This muscle is described with 8 strands originating from C3 to T3 and inserting into a common point of insertion on the occipital bone (Fig. A-12).

The longissimus cervicis is also a rather slender muscle which arises from the tips of the transverse processes of T1 to T5 just laterally to the longissimus capitis and inserts into the posterior tubercle of the transverse processes of C2 to C7. This muscle is described here with 5 strands with the longest arising from T5 and inserting in C2 and the shortest arising from T1 and inserting in C6.

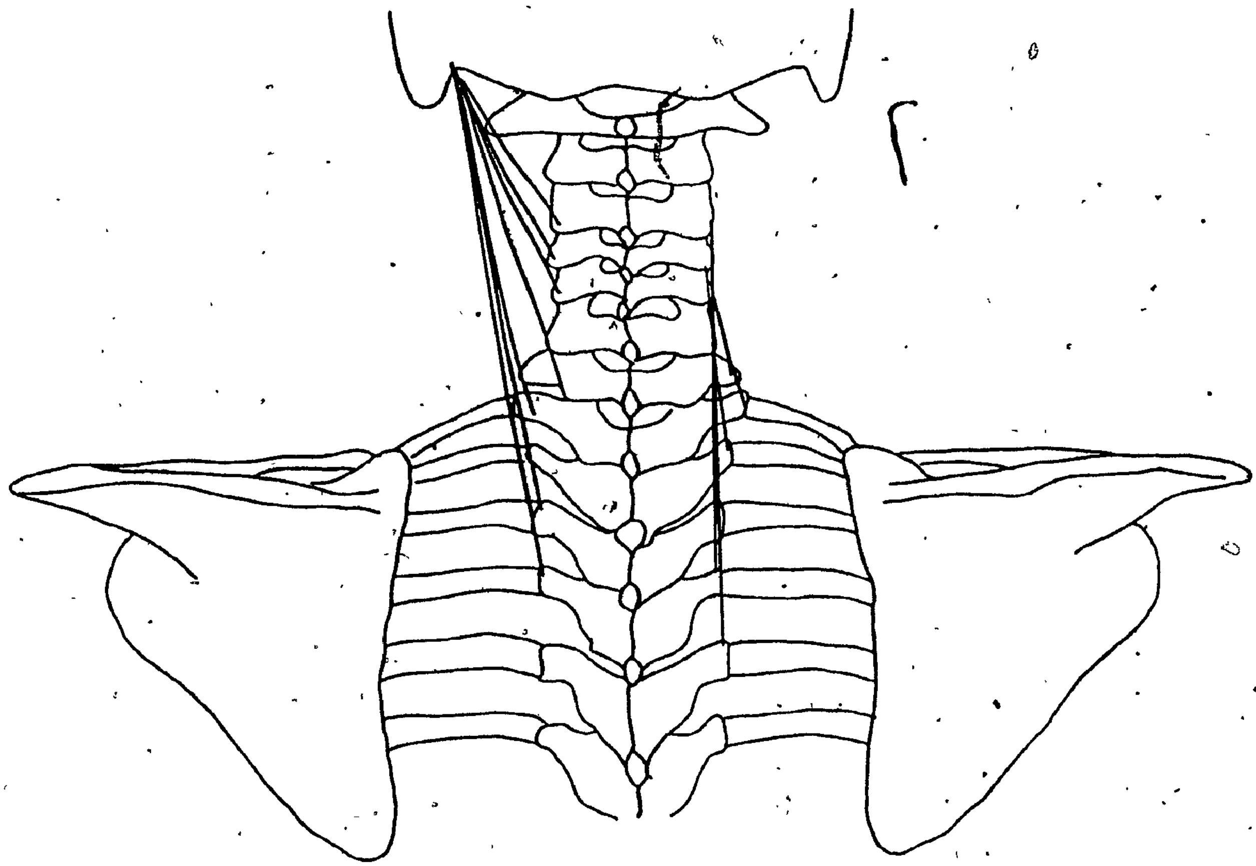
Calculations show that these two muscles extend and contract by a very small amount during flexion and extension. This result may be significant in determining the role that they play in supporting the head and neck.

The calculated shear, compression, and moment are relatively low and can be attributed to the rather small cross-sectional area of the muscle strands. The small moment produced suggests that their primary role may not be to support the neck during flexion.

Iliocostalis - Iliocostalis cervicis is a slender muscle much like the longissimus cervicis. It arises from the angles of the upper six ribs and divides usually into three slips of insertion that attach to the posterior tubercles of the transverse processes of C4 to C6.

This muscle is described by three pairs of strands. The strands originating from rib 1-2 insert into C6, the strands originating from rib 3-4 insert into C5, and the strands originating from rib 5-6 insert into C4 (Fig. A-13).

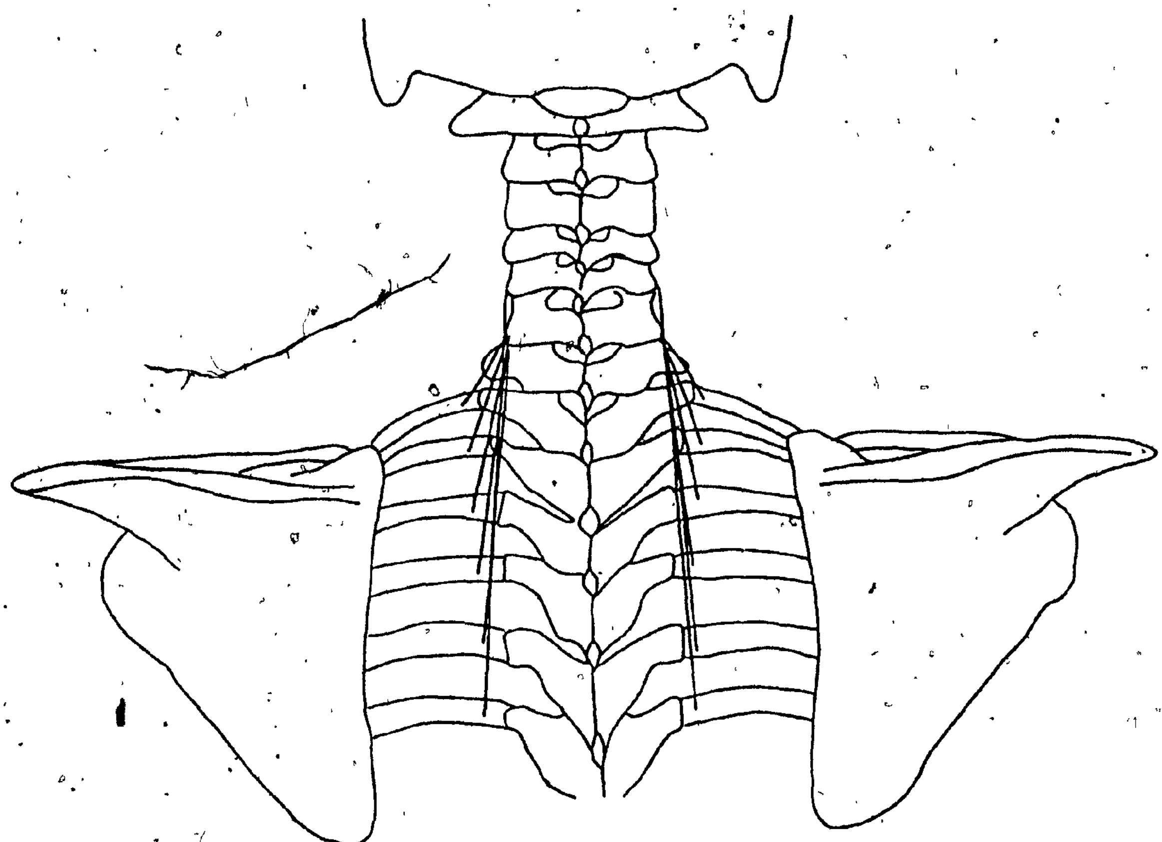
The muscle strands undergo little or no extension or contraction during flexion and extension and like the longissimus cervicis, the iliocostalis has a relatively small cross-sectional area. This



Muscle No. 6	Longissimus Capitis			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
55	601	408	109	0.3
56	602	508	109	0.3
57	603	608	109	0.3
58	604	708	109	0.3
59	605	808	109	0.3
60	606	907	109	0.3
61	607	1007	109	0.3
62	608	1107	109	0.3

Muscle No. 7	Longissimus Cervicis			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
63	701	907	707	0.3
64	702	1007	607	0.3
65	703	1107	507	0.3
66	704	1207	407	0.3
67	705	1307	305	0.3

Figure A-12 Vector description of the longissimus cervicis and capitis



Muscle No. 8 Iliocostalis

Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
68	801	1411	507	0.2
69	802	1311	507	0.2
70	803	1211	607	0.2
71	804	1111	607	0.2
72	805	1011	707	0.2
73	806	911	707	0.2

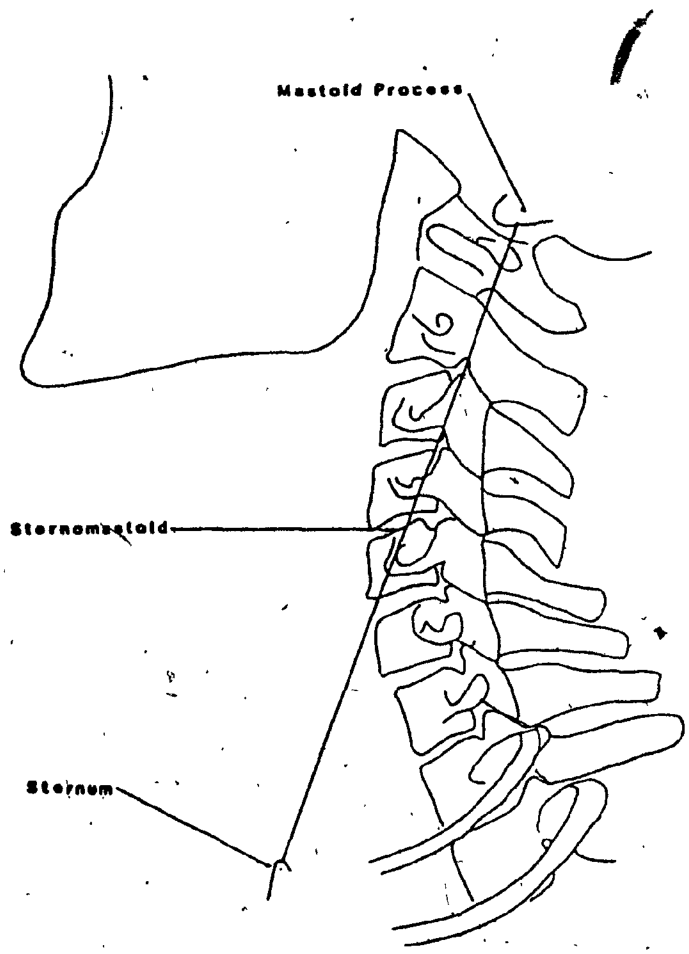
Figure A-13 Vector description of the iliocostalis muscle suggests that the role of the iliocostalis is the same as that of the longissimus cervicis.

Sterno-Mastoid - The sterno-mastoid is a long strong muscle which originates from the top of the sternum and the upper surface of the anterior portion of the clavicle and inserts into the mastoid process. This muscle is modelled with a single strand originating at the medial end of the clavicle and inserting into the mastoid process of the occipital bone (Fig. A-14).

The muscle contracts in both flexion and extension. This suggests that the muscle is at its maximum length somewhere in the vicinity of the neutral position. Another result worth noting is the change in the sign of the moment when going from the neutral to flexion position. This shows that the sterno-mastoid produces increased flexion when the neck is in flexion and increased extension when the head is in extension.

Scalenes - The scalene muscle group is made up of three muscles; scalenus anterior, scalenus medius and scalenus posterior. There is also a scalenus minimum which is considered to be part of the scalenus medius. Each of the scalene muscles is described separately although it is felt that they play the same role in supporting the neck and should probably be lumped together as one muscle. See Fig. A-15.

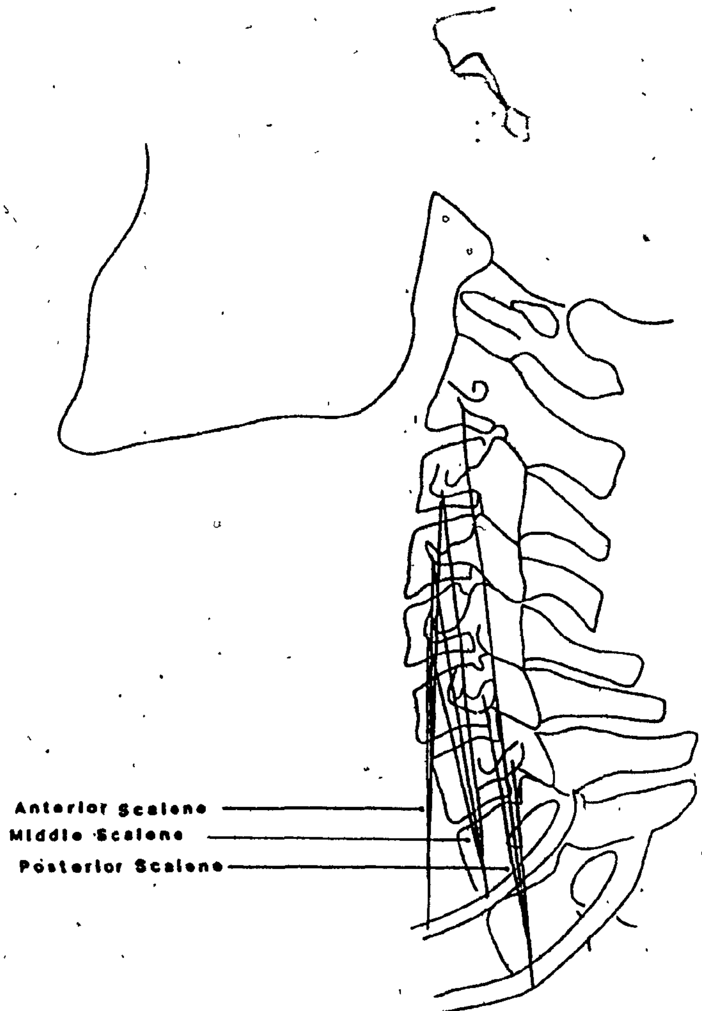
The scalenus anterior arises from the tuberculum scaleni of the first rib and inserts into the transverse processes of C3 to C6. This muscle is therefore modelled as 4 strands originating from a



Muscle No. 9 Sternomastoid

Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
74	901	1501	107	6.5

Figure A-14 Vector description of the sternomastoid muscle



Muscle No. 10	Scalene Posterior			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
75	1001	1603	607	0.9
76	1002	1603	707	0.9
77	1003	1603	807	0.9

Muscle No. 11	Scalene Medius			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
78	1101	1602	304	0.4
79	1101	1602	406	0.4
80	1101	1602	506	0.4
81	1101	1602	606	0.4
82	1101	1602	706	0.4
83	1101	1602	806	0.4

Muscle No. 10	Scalene Anterior			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
84	1201	1601	406	1.0
85	1202	1601	506	1.0
86	1203	1601	606	1.0
87	1204	1601	706	1.0

Figure A-15 Vector description of the scalene muscles common point of origin in the first rib and inserting into the transverse process of C3 to C6

The scalenus medius originates from the lateral surface of the first rib and inserts into the anterior points of the transverse processes of C2 to C7. Therefore this muscle is described as 6 strands all originating from a common point of origin on the first rib and inserting into C2 to C7

The scalenus posterior, which usually fuses with the scalenus medius, originates from the lateral surface of the second rib and inserts into the posterior part of the transverse processes of C5 to C7. Therefore the muscle is described as three strands originating from a common point of origin and inserting in C5 to C7.

The scalenus muscles do not change length appreciably during flexion and extension and they produce very little moment. These results could be expected from the fact that the muscles run almost in line with the body of the vertebra and have a small lever arm.

Longus - The longus muscle consists of the longus capitis and the longus cervicis, which may be divided further into superior, vertical, and inferior parts (Fig. A-16).

The longus capitis arises from the anterior tubercle of the transverse processes of C3-C6 and inserts into the lower part of the occipital bone.

The superior oblique part of the longus cervicis arises from the anterior tubercles of C3 to C6 and inserts into the anterior tubercle of C1. This portion of the muscle can be described by 4 strands arising from C3 to C6 and inserting into C1. The vertical portion arises from the bodies of C5 to T3 and inserts into the bodies of C2 to C4. This portion of the muscle is described by 6 strands with the longest running from T3 to C2 and the shortest running from C5 to C4. The inferior oblique portion arises from the vertebral bodies of T1 and T2 and insert into the anterior tubercles of C5 and C6. This portion is described with 2 strands, T1 to C5 and T2 to C6.

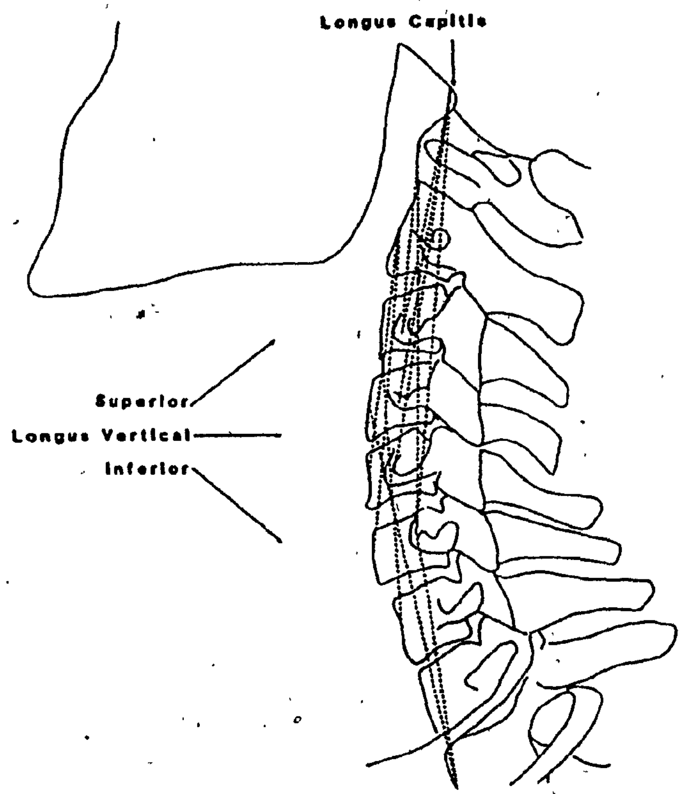
These muscles exhibit some change in length during flexion and extension. The moment produced by these muscles is relatively small. This small moment is due to a small cross-sectional area and to a short level arm.

Levator Scapula - The levator scapula arises from the vertebral margin of the scapula and inserts into the posterior tubercles of C1 to C5. The muscle is modelled by 4 strands arising from two points of origin on the scapula and inserting into C1 to C4 (Fig. A-17).

Calculations show a small change in length during flexion and extension, and a moderate moment due mainly to the large cross-sectional area of the muscle. This muscle behaves like the longissimus cervicis and iliocostalis.

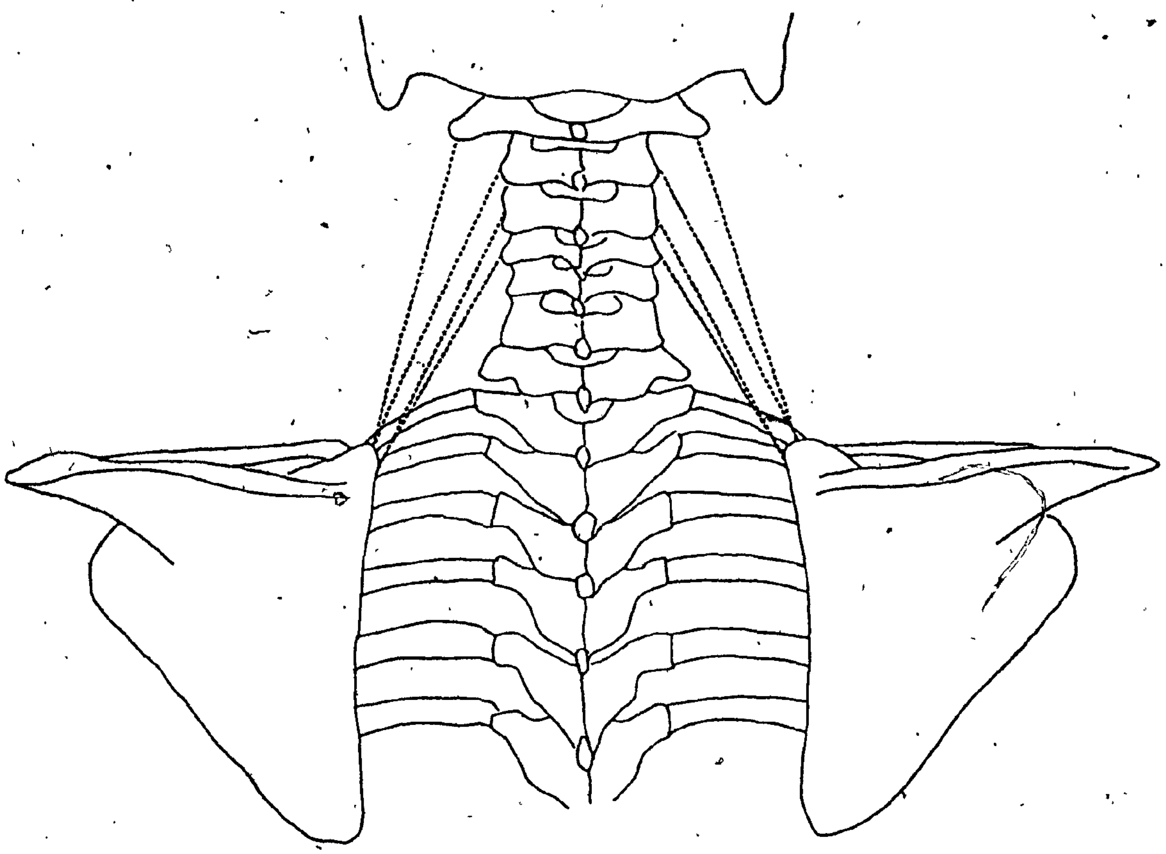
Note that the scapula is not necessarily a fixed body unless the arms are restrained. As stated before, the role of this muscle and other shoulder muscles depend upon the constraints applied to the arms. In this study the arms are assumed to be unconstrained and relaxed.

Trapezius - The trapezius is a broad flat muscle which originates from the posterior extremity of the clavicle, the acromion, and the



Muscle No. 13	Longus Capitis			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
88	1301	406	106	0.6
89	1302	506	106	0.6
90	1303	606	106	0.6
91	1304	706	106	0.6
Muscle No. 14	Longus Superior			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
92	1401	406	201	0.3
93	1402	506	201	0.3
94	1403	606	201	0.3
95	1404	706	201	0.3
Muscle No. 15	Longus Vertical			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
96	1501	605	505	0.1
97	1502	705	505	0.1
98	1503	805	405	0.1
99	1504	905	405	0.1
100	1505	1005	303	0.1
101	1506	1105	303	0.1
Muscle No. 16	Longus Inferior			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
102	1601	905	606	0.3
103	1602	1005	706	0.3

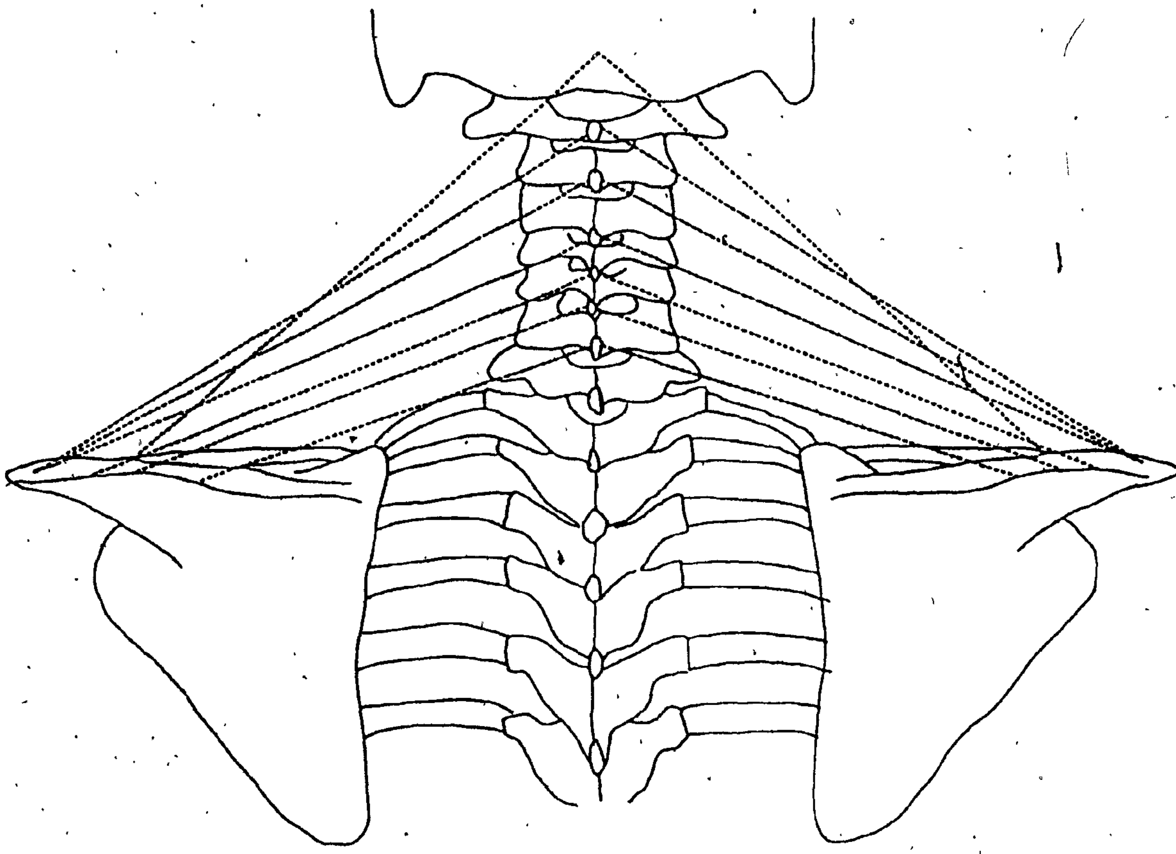
Figure A-16 Vector description of the longus capitis and cervicis



Muscle No. 17 Levator-Scapula

Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
104	1701	1511	203	1.3
105	1702	1511	305	1.3
106	1703	1510	407	1.3
107	1704	1510	507	1.3

Figure A-17 Vector description of the levator-scapula muscle



Muscle No. 18	Trapezius			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
108	1801	1502	103	1.3
109	1802	1503	204	1.3
110	1803	1504	308	1.3
111	1804	1505	411	1.3
112	1805	1506	511	1.3
113	1806	1507	611	1.3
114	1807	1508	711	1.3
115	1808	1509	811	1.3

Figure A-18 Vector description of the trapezius muscle spine of the scapula. The fibres arising from the clavicle insert into the ligament nuchae in the region of the occipital bone and the upper vertebrae. The fibres originating from the acromion insert into the ligament nuchae of the mid-cervical region and the fibres arising from the spine of the scapula insert into the remainder of the ligament nuchae and the ligament supraspinalis, as far down as T12 (Fig. A-18).

For this study, only the muscle fibres traversing the cervical joints are considered (i.e fibres inserting as low as C7). The muscle is modelled by 8 strands originating from eight points along the clavicle-scapula structure and inserting into the tips of the spinous processes of the seven cervical vertebra, and occipital bone. The three strands which insert into the occipital bone, Atlas, and Axis originate from three points along the posterior end of the clavicle. The two strands which insert into C3 and C4 originate from the acromion and the strands inserting into C5, C6, and C7 originate from three points along the acromion end of the spine of the scapula.

There is little change in length of the strands during flexion and extension. The trapezius produces large values for the moments, compression, and shear at all the joints.

Rhomboideus Minor - The rhomboid muscle consists of two parts, major and minor. Since only with the cervical joints are being considered and the rhomboid major inserts only into the thoracic region, only the description of the rhomboid minor is included in the model. This muscle originates from the vertebral margin of the scapula just below the levator scapula and inserts into the ligament nuchae of the spinous processes of C6 and C7 (Fig. A-19).

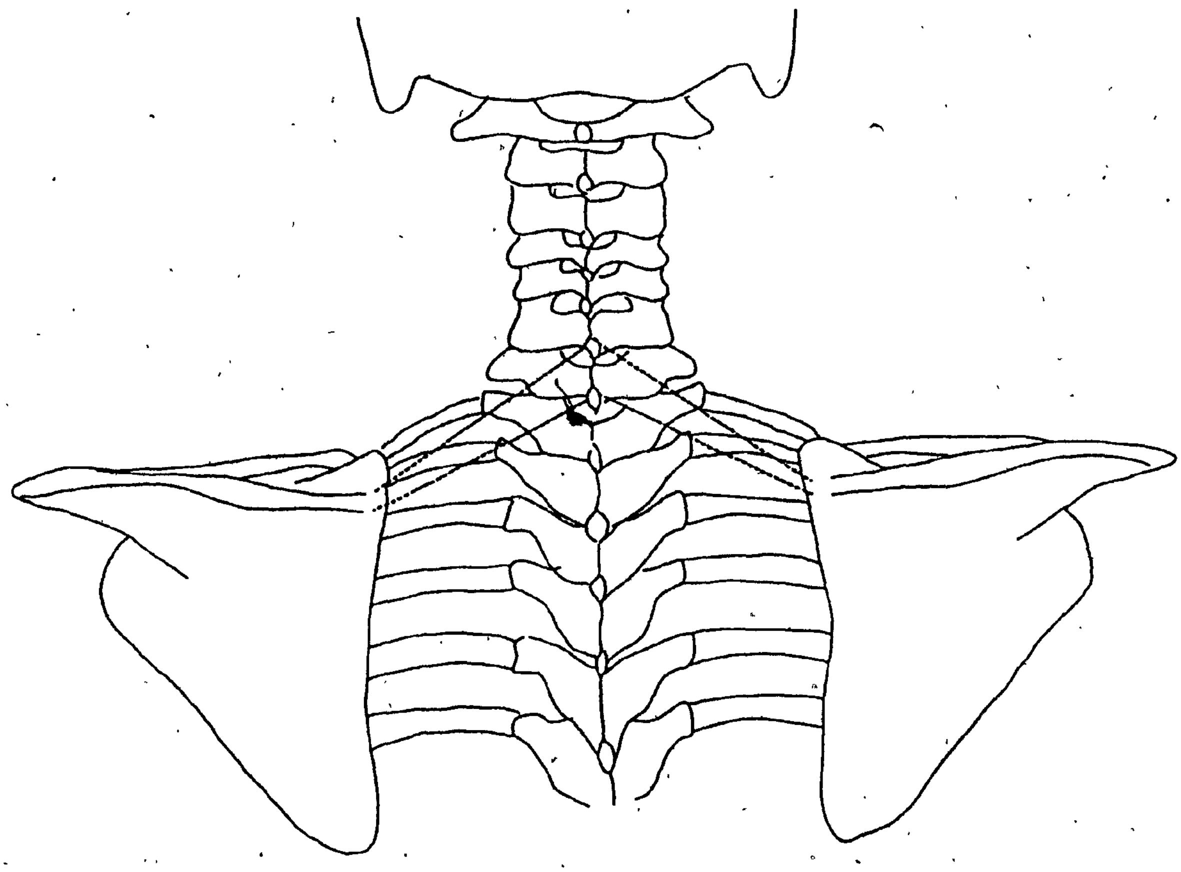
The muscle is modelled with 2 strands originating from 2 points of origin on the scapula and inserting into the tips of the spinous processes of C6 and C7. There is very little change in length during flexion and extension. This muscle produces a large moment due to its large cross sectional area and its long lever arm.

Like the other muscles originating from the scapula, it is assumed that it can produce a force only when the arms are restrained.

Serratus Posterior Superior - Originates from the 2nd to 5th rib and inserts into the ligament nuchae and the spinous processes from C6 to T2. Only the upper half of this muscle which inserts into C6 and C7 is being considered. Therefore it is modelled as 2 strands originating from rib 2 and 3 and inserting into C6 and C7 respectively (Fig. A-20).

This muscle behaviour is similar to the one calculated for the rhomboid minor. The change in length is small and the lever arm is large. The magnitude of the force produced by this muscle is smaller than those produced by the rhomboid due to the smaller cross-sectional area.

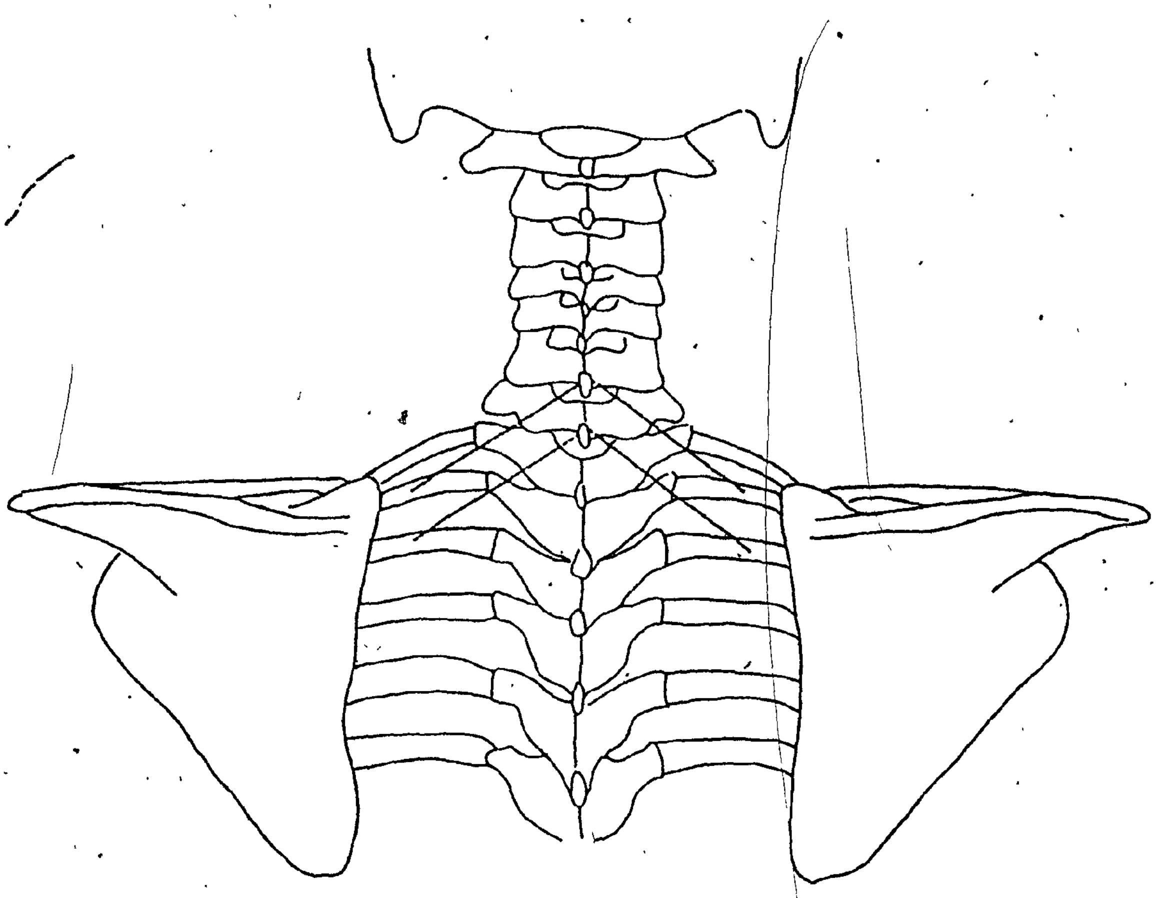
The serratus posterior superior and the rhomboids may not play a major role in supporting the neck but may be used strictly to support the ribs and the scapula respectively.



Muscle No. 19 Rhomboid

Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
116	1901	1512	711	2.6
117	1902	1513	811	2.6

Figure A-19 Vector description of the rhomboideus minor muscle



Muscle No. 20	Serratus Posterior Superior			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
118	2001	1604	711	0.3
119	2002	1605	811	0.3

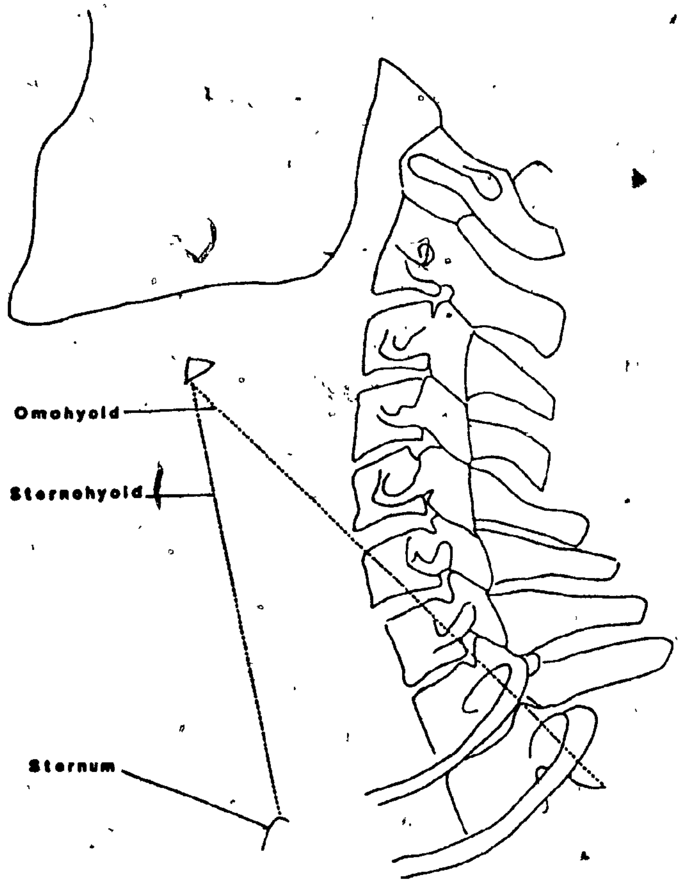
Figure A-20 Vector description of the serratus posterior superior

Sternohyoid and Omohyoid - These anterior muscles of the neck have a more complicated arrangement than the muscles described so far. They run from the hyoid bone to the occipital bone and the jaw, and also run from the jaw to the head. The combined action of these muscles can produce a forward bending moment on the head as well as compression and shear (Fig. A-21).

In order to simplify the analysis, these muscles are divided into two groups: those superior to the hyoid bone and those inferior to the hyoid bone. It is assumed that the muscles above the hyoid are capable of producing a force at least as great as the force produced by the muscles below the hyoid bone. This allows for simplification of the model by considering the forces produced by the muscles below the hyoid bone to act directly on the head. The assumption that the muscles above the hyoid bone produce a sufficient pull to balance the muscles below can be easily verified by measuring the cross-sectional area of the two groups of muscles and considering their lines of action.

The muscles which run down from the hyoid are the thyrohyoideus, the sternothyroid, the sternohyoid and the omohyoid. The first three of these muscles produce a line of action running from the hyoid bone to the sternum and are grouped together and named the sternohyoid muscle. The omohyoid muscle originates from the superior margin of the scapula and inserts into the hyoid bone.

The two muscles are modelled as single strand muscles originating from the scapula and the sternum and inserting into a common point of insertion on the hyoid bone. The sternohyoid shows a decrease in length of 56% due to flexion. This would appear to be an unaccept-



Muscle No. 21	Sternohyoid			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2)
120	2101	1501	1701	2.4

Muscle No. 21	Omohyoid			
Strand No.	ID. No.	Origin	Insertion	Area(cm^2).
121	2201	1514	1701	0.6

Figure A-21 Vector description of the sternohyoid and omohyoid muscles able value for muscle contraction, and this suggests that the muscle is going slack. The sternohyoid also produces a very large forward moment due to its long lever arm whereas the omohyoid produces a much smaller moment and hence may not play a large role in supporting the load on the head and neck.