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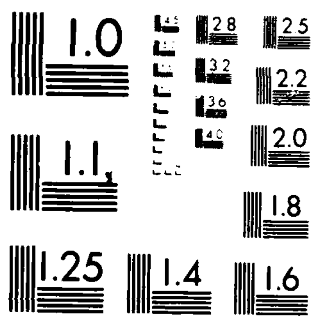

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CERVICAL SPINE ANALYSIS FOR EJECTION INJURY PREDICTION

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CERVICAL SPINE ANALYSIS FOR EJECTION INJURY PREDICTION

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ABSTRACT

We have developed a sagittal plane mathematical model for the cervical spine (including T6-T1, C7-C1 and skull). In our 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 muscle, ligament, and intervertebral joint. With this formulation, the problem is to find a method for distributing the moment between muscle and ligament.

Each of the possible solutions was graded against a mathematical objective function containing the equality constraint (i.e. moment must be balanced); the inequality constraints (finite limit to muscle force, finite strength of joint components); and the stress experienced by each joint. We then selected the unique solution that produced a minimum of stress at the intervertebral joints.

The model has been validated by human experimentation, in which volunteers were asked to exert a voluntary pull with their head against a variable resistance. Electromyographic measurements of various superficial neck muscles have been matched with predicted patterns.

Our calculations show that the mathematical representation of physiological behavior demands that stress be minimized at the intervertebral joint. It is interesting to note that Wolff has observed that bone architecture at the microscopic level responds to stress. Our findings suggest the system as a whole is controlled by stress.

This validated model was then subjected to simulation in order to determine the maximum acceleration that the cervical spine would take for different postures. We 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. The worst case was calculated to be 13 g's and the best case 37 g's. 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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CERVICAL SPINE ANALYSIS FOR EJECTION INJURY PREDICTION

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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, and 5) the evaluation of the relative efficiency of various surgical and non-surgical corrective techniques of the spine, let alone a method for spine evaluation, and a technique for matching a spine to a specific type of work.

I- REVIEW 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.1.1). 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 transverse 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. 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.

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, and are necessary to satisfy the conditions for geometrical fit.

Fresh human cadaver cervical spine specimen are rare and reports generally are based on small numbers. The resistance of the cervical spine joints to compression, lateral bend and torsion have been published [6]. These tests indicate that the behavior of these joints and of the vertebrae resembles the behavior of their equivalents in the lumbar spine [23, 12, 43, 16, 4, 7, 37, 9, 30, 26, 11]. The special arrangement of the axis/atlas joint has been studied by Panjabi [33].

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, or of such structures as the atlanto-occipital membrane, the anterior and posterior longitudinal ligaments of the neck, or the cruciate ligaments.

Motions of the cervical spine joints have been reported in the literature [15,13,1,8] and their numerical values correspond closely. The pattern of cervical spine motion has been beautifully 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.

The electromyographic studies, which can be used as a timing device for muscle activity, are also only sparsely reported in the literature. The normal erect posture is defined by Basmajian [5]. He also presents the best summary of EMG studies. The muscles of the neck are sparingly mentioned; those of the shoulder get more attention but particularly with respect to the upper extremity motion. Other EMG studies describe the activity of longus colli and longissimus cervicis [14]. These EMG studies do not give integrated outputs. Such outputs have proven very useful in our lumbar spine model [18].

The force of muscle contraction has been estimated to be 50-120 psi [20]. We have shown that it is possible to estimate muscle moments using sectional anatomical specimens to derive muscle areas [10,11]. This procedure has been supported by Rab and Chao [36] who made almost identical estimates.

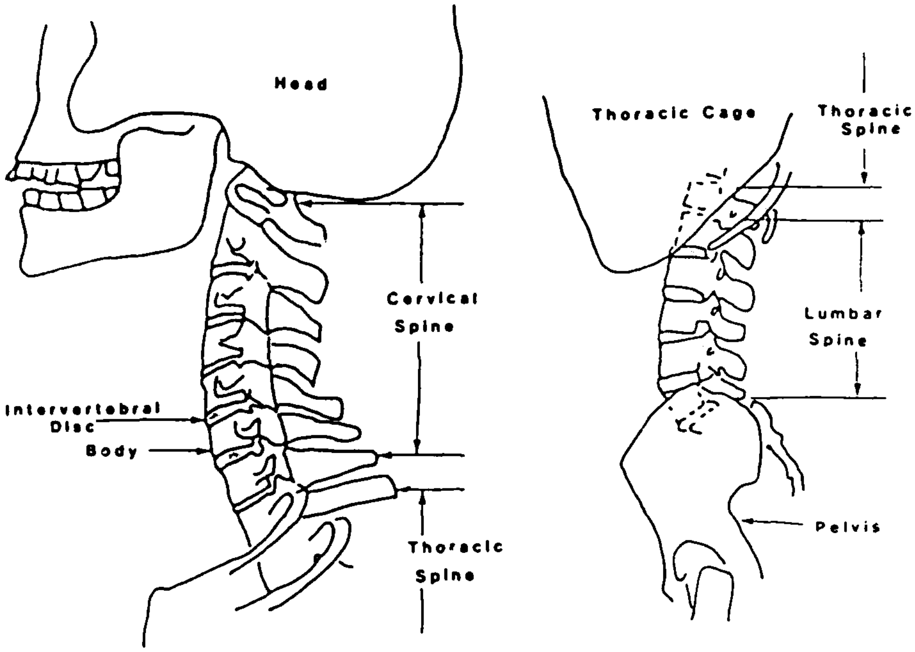

Figure 1.1 Lateral view of the cervical spine and the lumbar spine

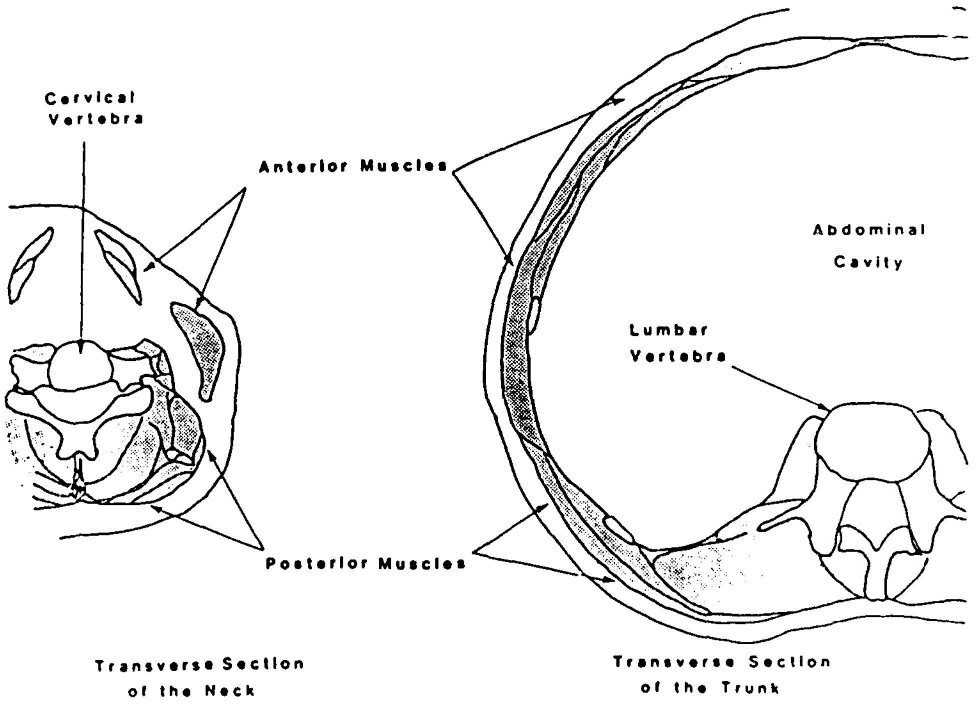

Figure 1.2 Transverse section through the neck and trunk

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. The anterior and posterior muscle mass of the neck and trunk are illustrated in Fig.1.2, and the muscular structure associated with the lumbar spine is illustrated in Fig 1.2.

II- REVIEW OF MODELLING APPROACHES

The existing modelling approaches have been divided into three categories:

1) the continuum model
2) the discrete parameter model
3) the optimization model

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 optimization models may be a priori considered in the same group with the discrete parameter models as far as the mathematical description of the spine is concerned. However the utilization of such a model is sufficiently different to consider them in a separate group.

1- THE CONTINUUM MODEL

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 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 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.

2- THE DISCRETE PARAMETER MODEL

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 Gx 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 has been 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 acceleration in excess of five g's 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.

3- THE OPTIMIZATION 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. In the Arvikar and Seireg [3] model no attempt is made to account for the difference in the joint's ability to support the loads in the two main directions.

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

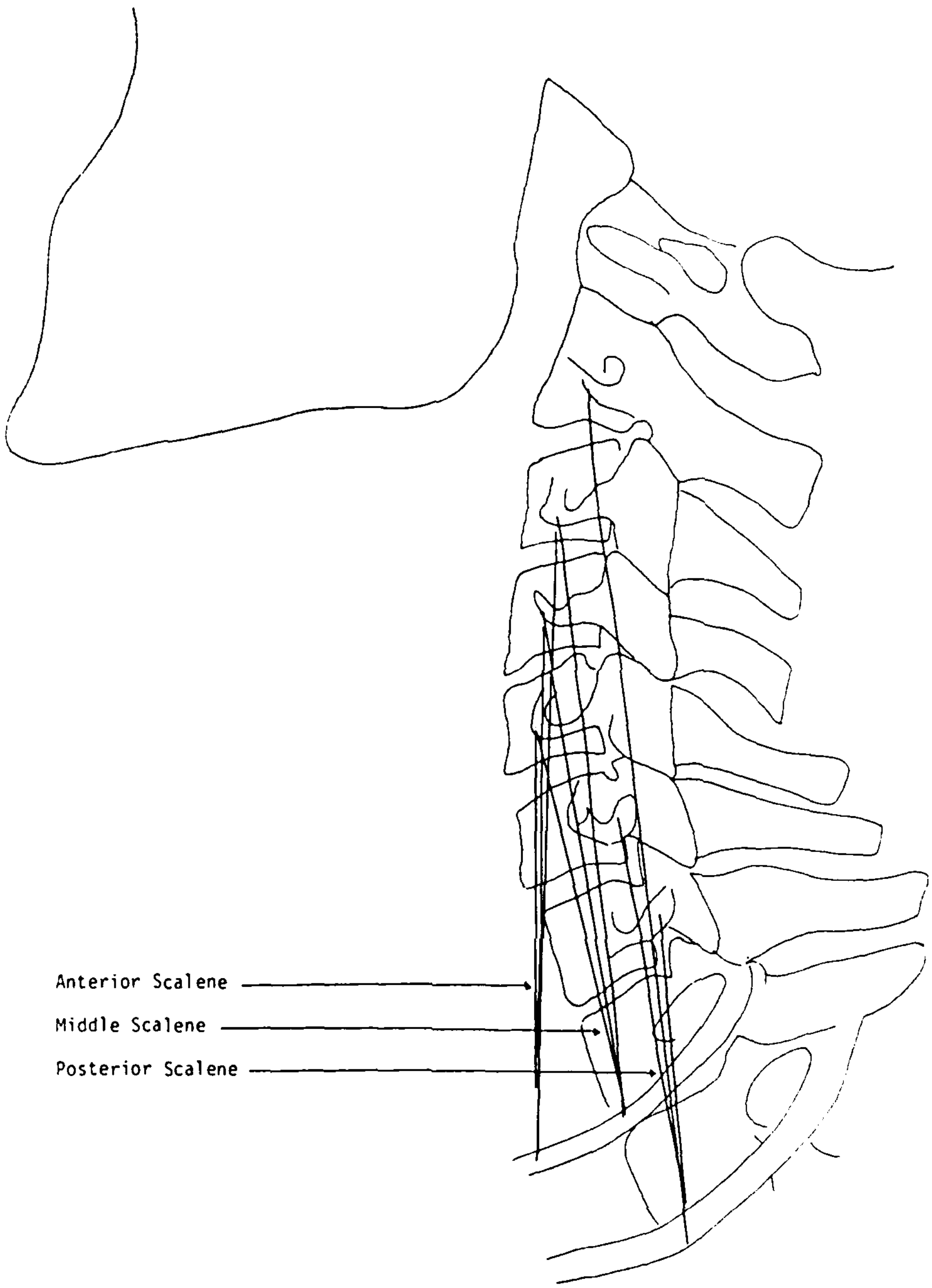

Figure 3 Vector representation of Scalene muscles

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:

$$F = K + C1*M + C2*R$$

where K = muscle force
M = reaction moment of the joint
R = reaction force on the joint
C1 and C2 are weighting factors

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 a subject was modelled 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 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.

We can infer from Wolff's law that the stress is equalized throughout the vertebra. Because bone tissue is identical regardless of the spinal level, it follows that the stress in all spinal skeletal segments 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 elaborate 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 we should do 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 we replaced the distributed forces at the vertebral end plates by a resultant vector. Similarly the muscles and ligaments were represented as a finite set of vectors with points of origin and insertion.

For reasons of symmetry we have assumed that the center of reaction of two adjacent vertebrae is in the sagittal plane and lies in the bisector of the disc. There exists a certain amount of experimental evidence to support that assumption.

A simplified model of the lumbar spine that requires the stress to be minimized could be validated in two ways:

1- By obtaining a solution that will be 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 will explain 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.

We have written the equations representing a model of the spine incorporating all the above features in the case of a weightlifter executing a dead lift [18]. This model was indeed validated by the fact that it explained all published experimental data on the lumbar spine. The model further predicted that the optimal lifting sequence is achieved when the stress is equalized at all intervertebral joints. The objective function used in the optimization is of the following form:

$$F = P1*K + P2*S + P3*C + P4*M$$

where K = Sum of the squares of the muscle firing density
S = Sum of the squares of the shear at each IV joint
C = Sum of the squares of the compression at each IV joint
M = Sum of the squares of the reaction moments at each IV joint
P1,P2,P3 and P4 are weighting factors

This validated model has proven 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.

In short, the advantages of the optimization method over the previous approaches are 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.

EMG INVESTIGATION OF THE HUMAN CERVICAL SPINE

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 our 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.

PRELIMINARY INVESTIGATION

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

The procedure followed was 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 was placed behind the earlobe. A colloidian glue was used to facilitate electrode adhesion in the hairy regions of the neck (ie. semi-spinalis capitis and splenius capitis).

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

Electrode placement - A number of muscles were monitored for EMG activity in a number of loading configurations. In all cases, EMG recordings were 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) **Semispinalis Capitis** The placement of the electrodes for this muscle is 2cm below the occipital bone and 2cm lateral to the midline (8). It was found that activity could 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 was found that activity could be observed from this muscle during extension against resistance and rotation against resistance.

3) **Sternomastoid** The electrode placement for this muscle was chosen at 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 was observed during rotation and flexion of the head against resistance. Little or no activity was observed during extension against resistance.

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

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

6) **Sternohyoid** A single pair of electrodes was placed slightly below the hyoid bone. Activity could be observed during flexion against resistance and no activity was recorded during extension 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 : RESULTS FROM PRELIMINARY EMG INVESTIGATION

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 will be monitored. Based on the results described above, it was 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 were 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 consisted 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 re*sultant load. It is important* to 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 were examined (10 males and 7 females) ranging in age from eighteen to seventy-nine. Paired miniature silver-silver chloride electrodes were placed bilaterally on the sternomatoid, splenius capitis, and omohyoid muscles. The semispinalis capitis muscle was 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 required verification. A simple opening of the jaw proved to be sufficient for this purpose.

The loading apparatus consisted of a typical strain gauge device mounted horizontally with a comfortable sling attachment to accommodate the head. Seated in a chair, the subject was asked to pull back against the sling, thereby increasing his extensor moment (Fig.2.1). The shoulders were relaxed with the arms hanging loosely at the side. An idealized view of the loading arrangement is depicted in Fig. 2.3.

A sequence of photographs was taken of each exercise at 4 frames per second. Onto these, lines were 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 was possible to detect any change in geometry during the performance of the task.

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.2.2) 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 were calculated at 25% and 4% of their assumed maxima respectively.

In all 17 subjects the following observations were 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.2.4 and Fig. 2.5). It was 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 was pulling.

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

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

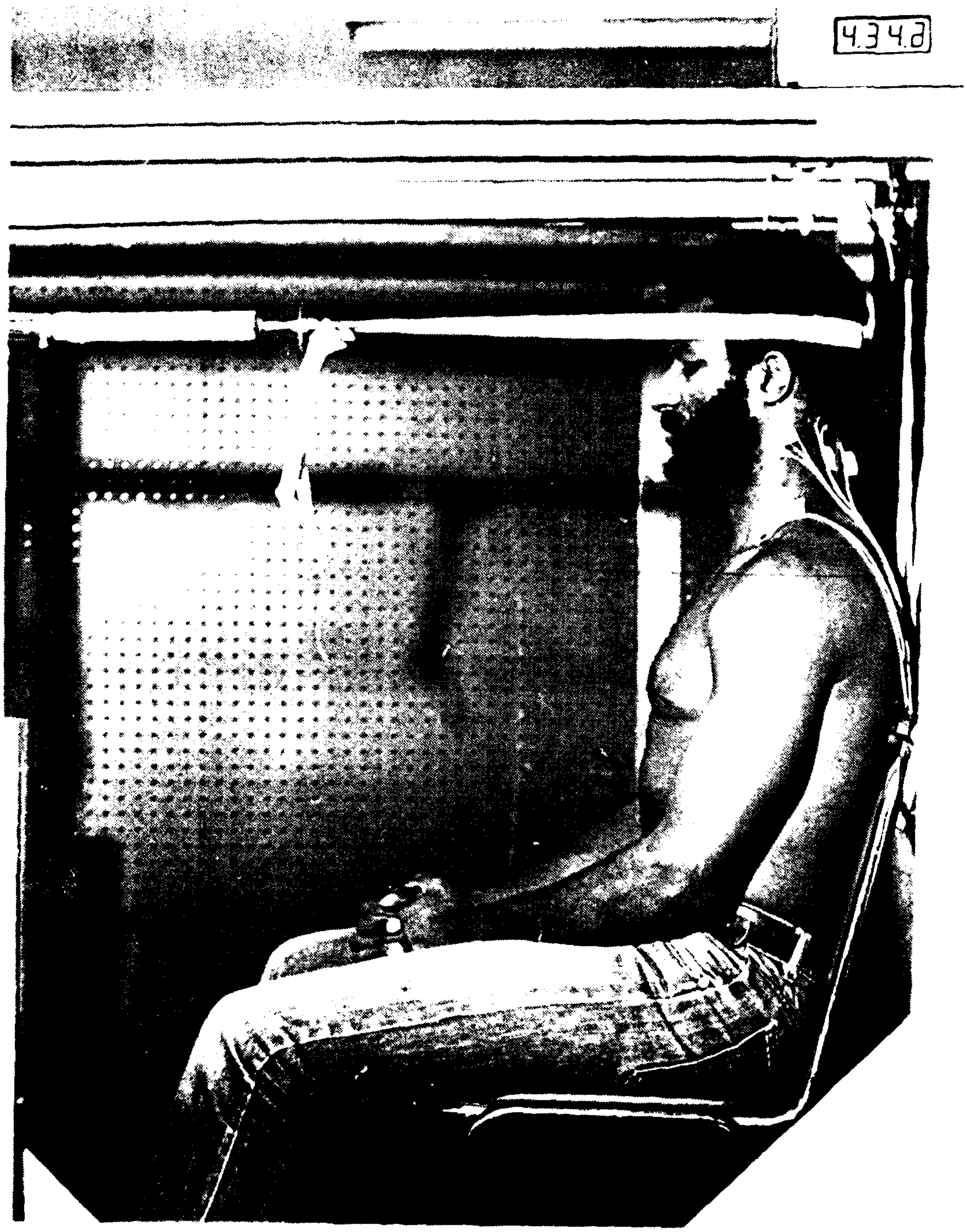

Figure 2.1 Lateral view of experimental loading procedure

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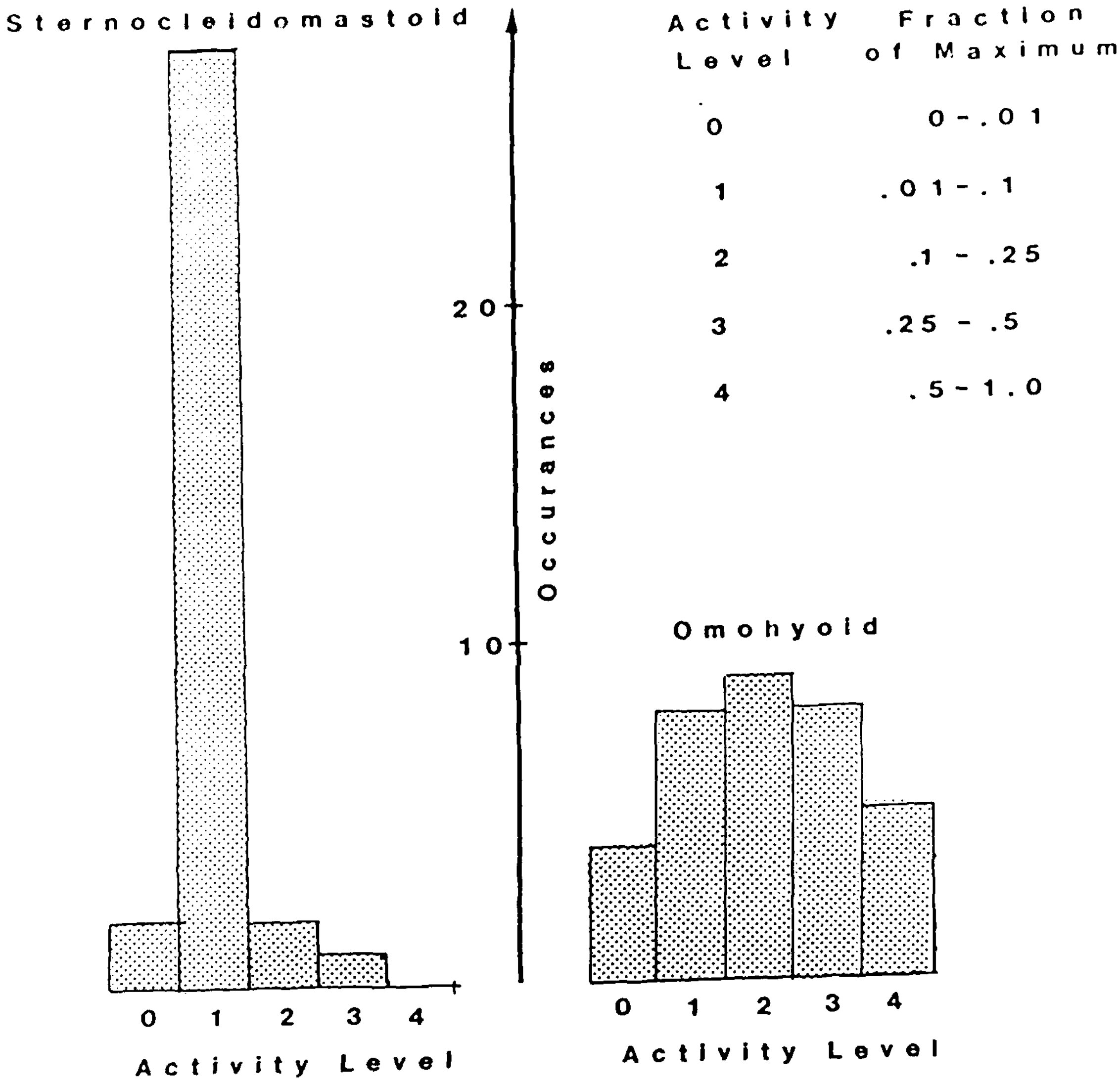


Figure 2.2 Results for sternomastoid and omohyoid muscles

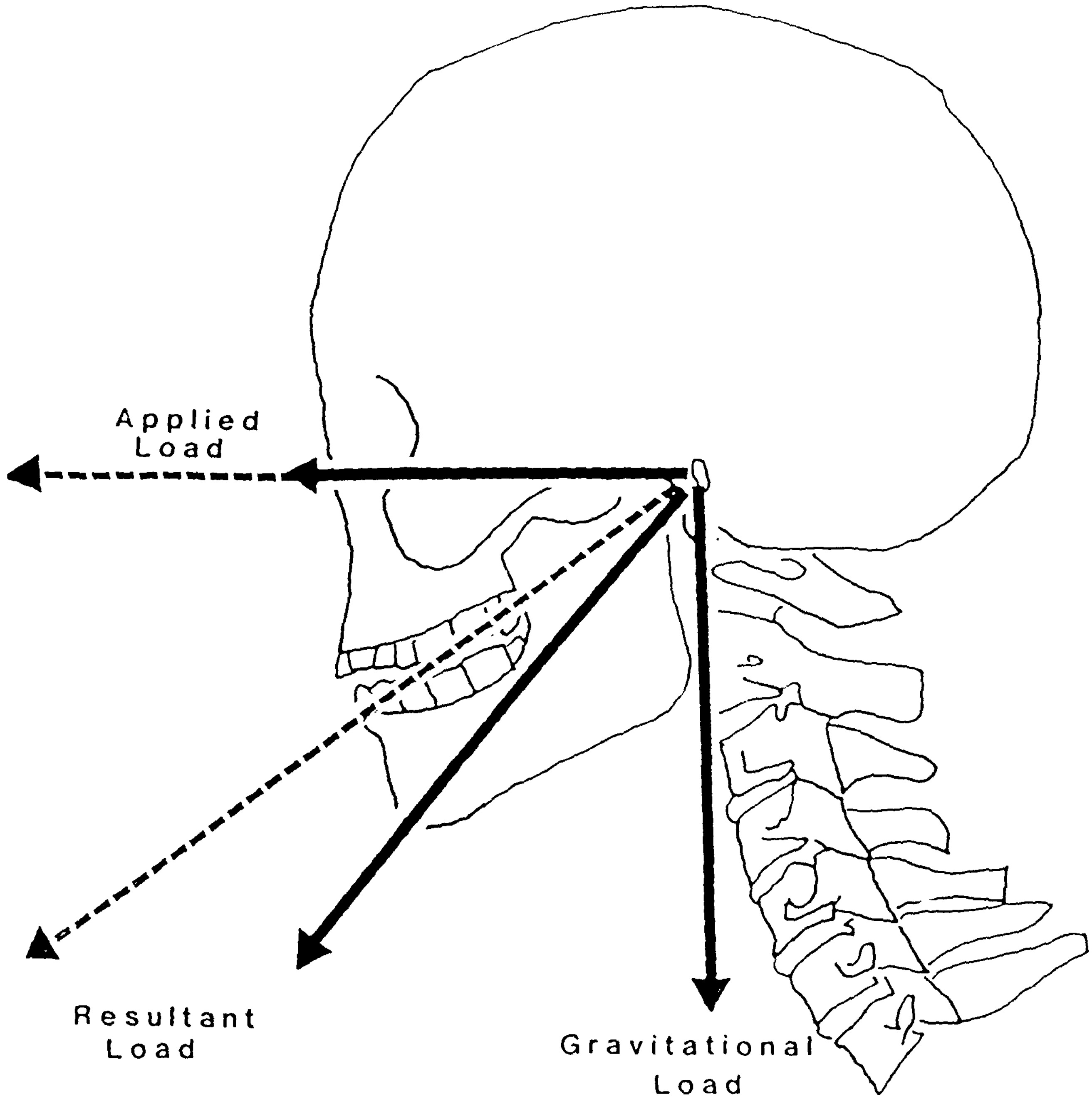


Figure 2.3 Idealized view of experimental loading procedure

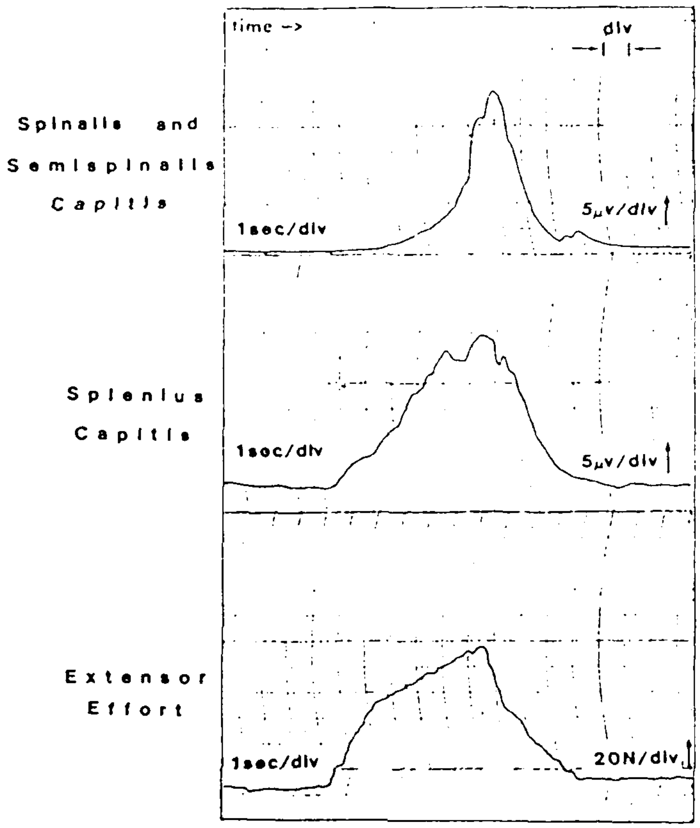


Figure 2.4 Sample IEMG for semispinalis capitis and splenius capitis

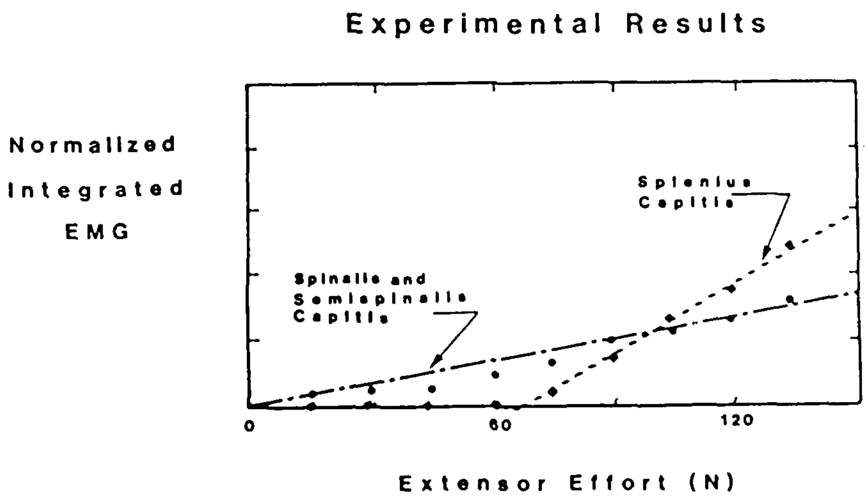


Figure 2.5 Experimental muscle firing pattern for semispinalis capitis and splenius capitis muscles

MAIN INVESTIGATION

CHANGES TO THE APPARATUS The load cell and the load cell mounting apparatus were changed as follows:

1) The strain gauge rod was 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 was provided by a strap which passed around the subject's head and attaching to the strain gauge bolt.

3) The strain gauge was 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 was bolted to the same rigid frame which was 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 was asked to attempt to pull his head backwards against the restraining strap. This produced a flexion load which is monitored with the load cell apparatus. The analog signals resulting from the load cell and the amplified electromyographic signals were 1) analog filtered in the band 10Hz to 250 Hz and 2) digitized (12 bits) at a rate 1000 samples per second and stored on digital disk. The analog filtering adequately attenuated the biopotential signal resulting from the heart and eliminated high frequency noise.

The test performed on each of the volunteers consisted 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 could 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 were removed and replaced bilaterally over the posterior scalene muscle and unilaterally over the trapezius and sternohyoid muscles. The procedure of task 1 is repeated.

Five volunteers were 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 was 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.

RESULTS

The results of the preliminary investigation suggest that the information which is the most useful in validating 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 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. 2.6. The corresponding RMS value is plotted versus time in Fig. 2.7. 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.

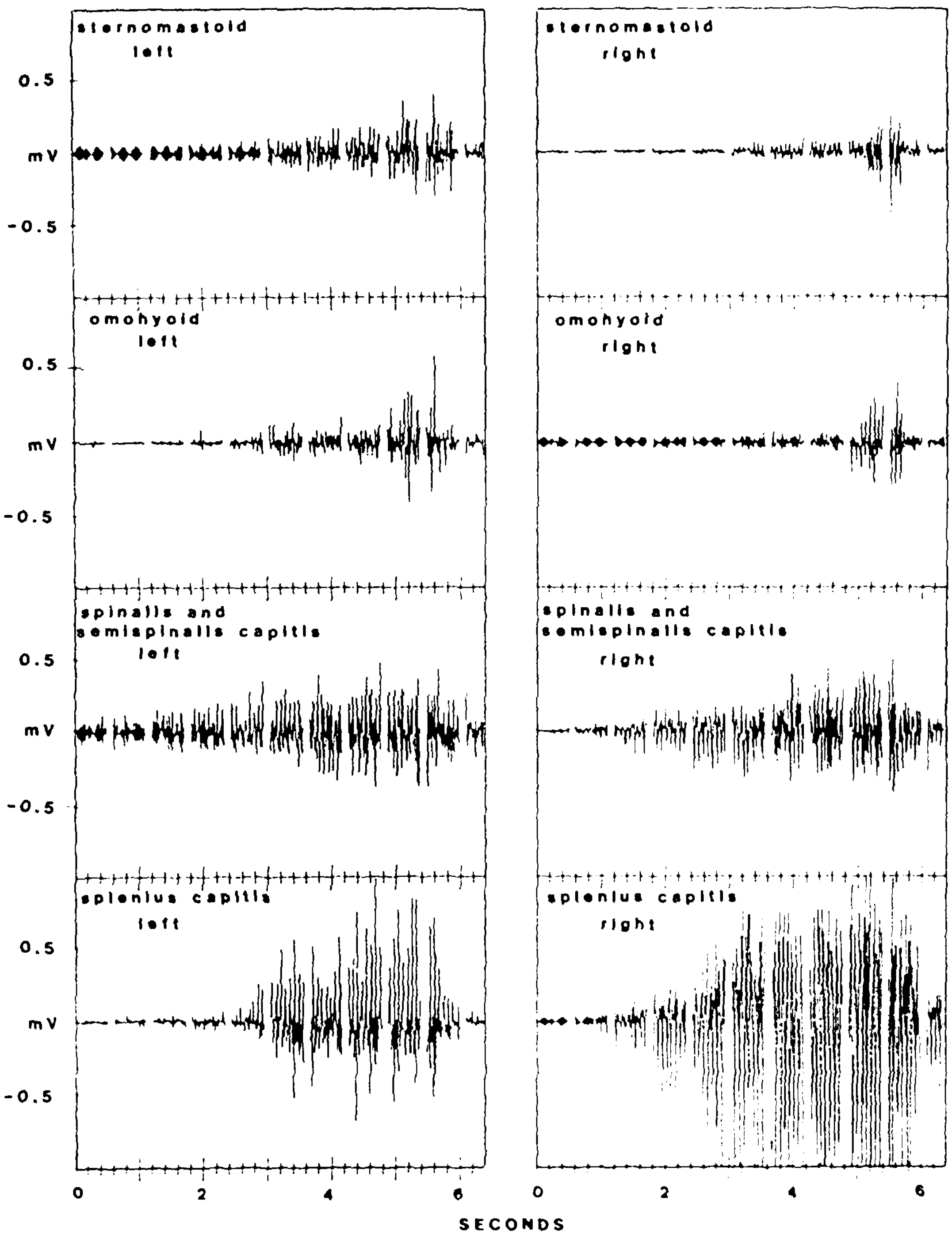


Figure 2.6 Sample EMG data for volunteer #1 performing TASK - 1

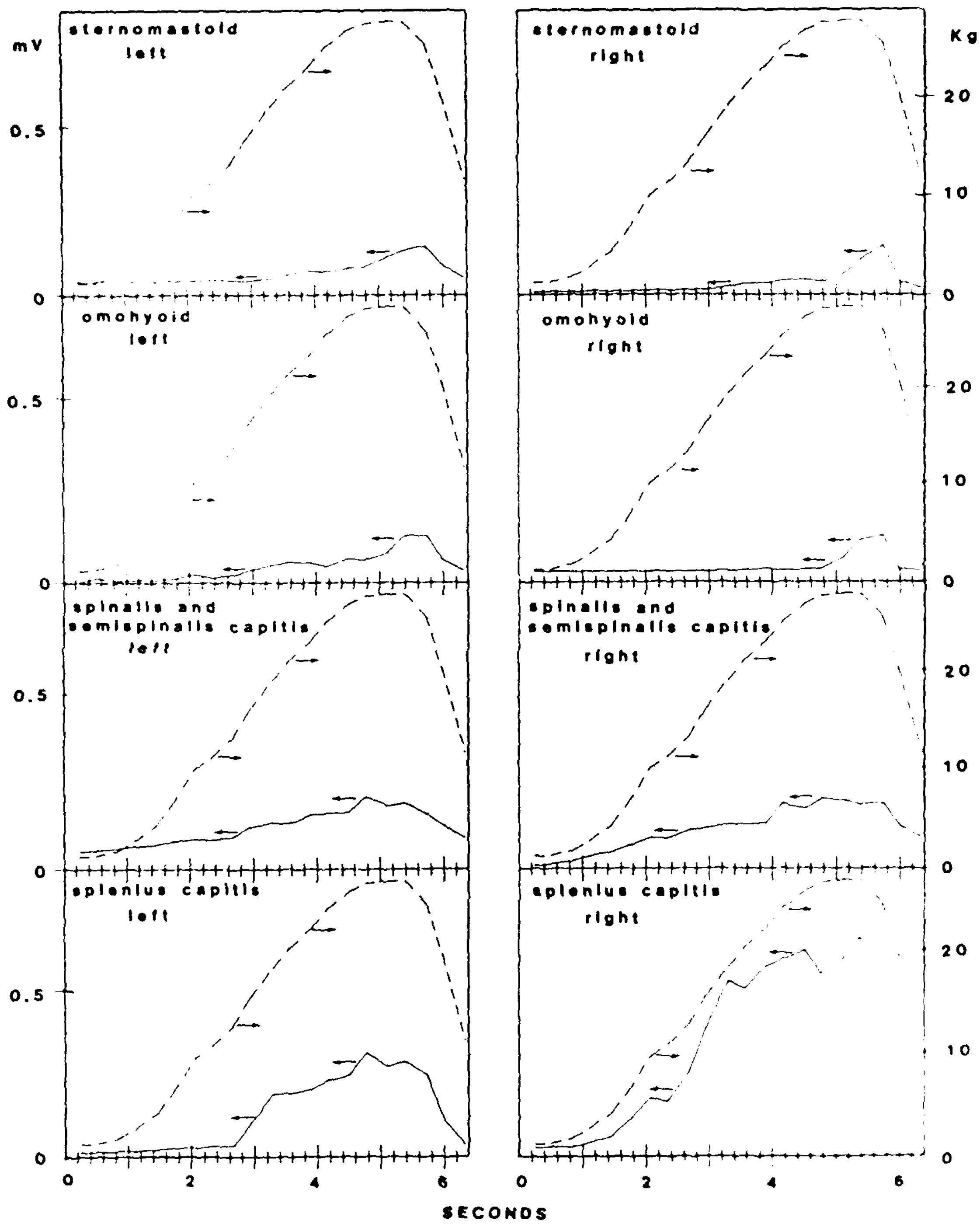


Figure 2.7 Sample IEMG results for volunteer #1 performing TASK - 1

Variations were observed in the results between the different volunteers, between tests performed on the same volunteer (as illustrated by the sample results of Fig. 2.7), between results from the muscle mass on opposite sides of the spine for the same muscle during the same test. It was 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

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. 2.8). 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 semispinalis capitis and splenius capitis muscles are recruited in two distinctly different ways. As the extensor moment is increased, the semispinalis capitis muscle exhibits an immediate response whereas the splenius capitis does not. This delay was 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 was not observed, since muscle activity as a function of the extensor effort was 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.

The study has shown the omohyoid muscle to be active during extension of the head against resistance. Because of the unlikely position of the muscle, these findings are 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 showed a wide spread of values. It was not 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.

TASK	VOLUNTEER	MAXIMUM: EFFORT (kg)	MUSCLE ACTIVITY ONSET: STERNOMASTOID (kg)	OMOHYOID (kg)	SPINALIS CAP (kg)	SPLENIUS CAP (kg)
	1	25.8	10.1	16.0	0.0	5.7
	2	25.4	13.2	12.5	2.4	7.5
1	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
	1	22.5	5.3	9.7	1.1	3.0
	2	20.0	14.1	16.1	3.6	4.8
2	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
	1	30.4	8.2	11.2	0.0	8.5
	2	27.0	8.0	18.8	0.4	10.9
3	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
	1	20.1	9.5	13.0	2.7	4.2
	2	17.8	6.6	12.4	0.6	5.1
4	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
	1	32.0	10.2	14.0	0.0	7.0
	2	29.3	16.3	16.3	0.0	9.0
5	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 1-5

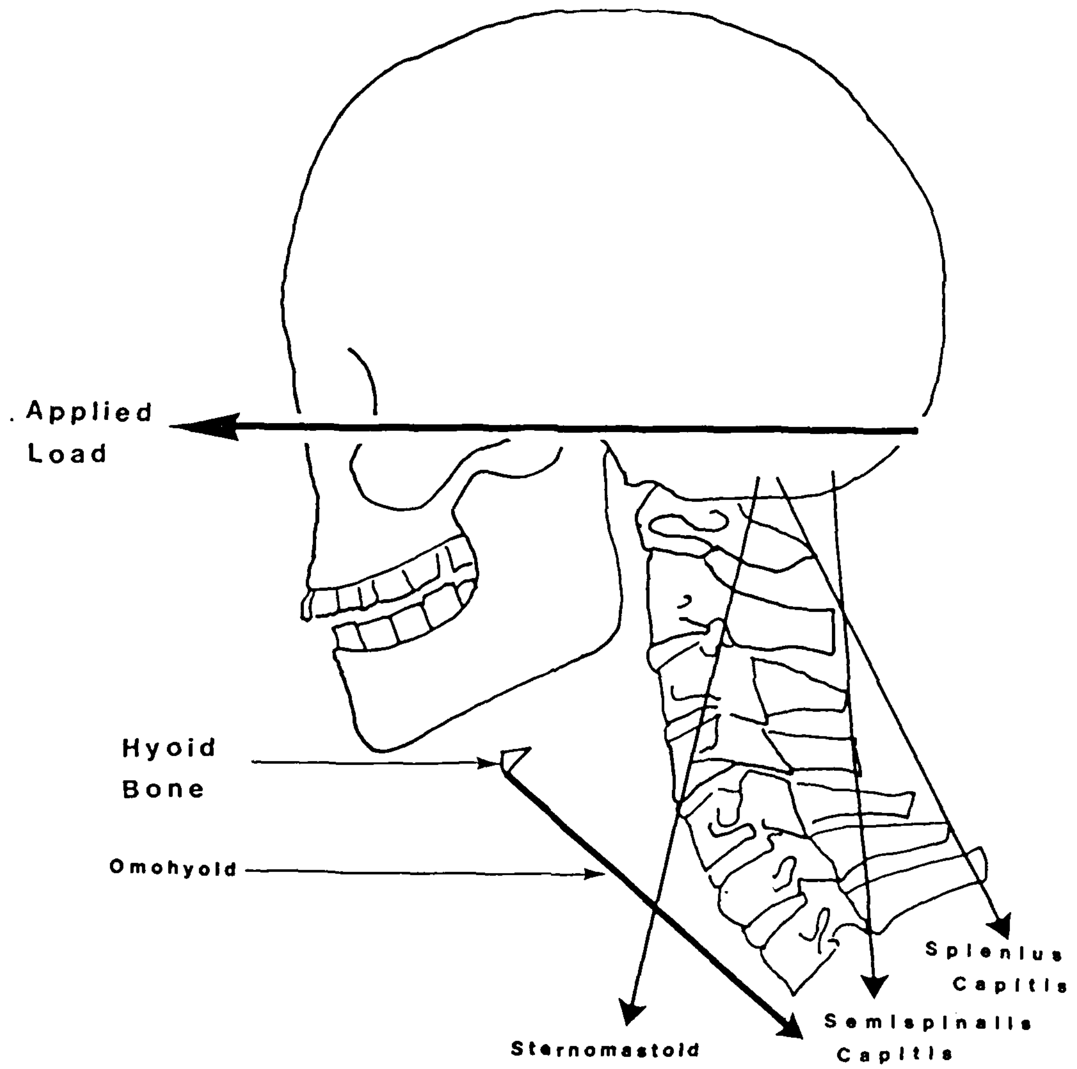


Figure 2.8 Average force produced by muscles of interest

Intuitively, we can see from Fig.2.8 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 was 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. 2.8 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 were also 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.

MODELLING THE HUMAN CERVICAL SPINE

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 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 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 were used in the development of the lumbar spine model will be used to develop a model of the cervical spine.

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, we will consider a single cervical joint as shown in Fig. 3.1.

In this illustration, we see the joint supporting 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.

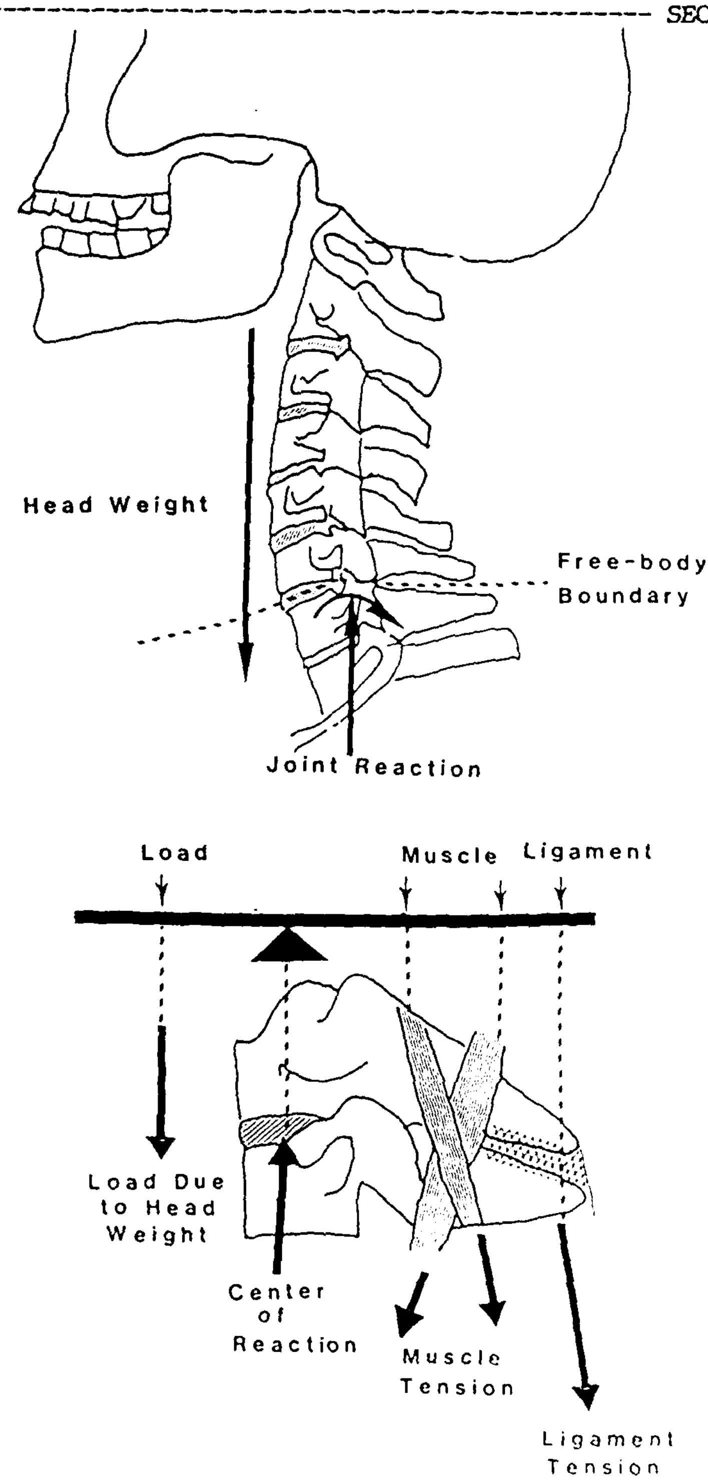


Figure 3.1 Simplified free-body analysis of C6-C7 intervertebral joint showing load distribution in cervical joint

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. 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 possible to impose additional constraints on this system and still satisfy the equilibrium conditions.

Wolff's law, as stated elsewhere in this report, characterizes the response of the bone to stress. We have 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 result in stress minimization and equalization at each intervertebral joint.

WOLFF's LAW APPLIED TO THE CERVICAL SPINE.

OPTIMIZATION

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

DEFINITION : THE OBJECTIVE FUNCTION

The objective function expresses the necessity for the intervertebral joints to minimize and equalize their stress without exceeding physiological limitations (limitations of muscular power and the strength of ligaments). The stress is due to the actions of the muscle pull (Kk), the ligament reaction (Lj) and the shear reaction of the joints (Jj). This objective can be approximated as a quadratic function as follows:

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

where:

$$F_1(\text{muscles}) = \underset{k=1\text{toNm}}{\text{SUM}} (P1xK_k)^2$$

$$F_2(\text{shear})= \underset{j=1,8}{\text{SUM}} (P2xJs_j)^2$$

$$F_3(\text{comp.})= \underset{j=1,8}{\text{SUM}} (P3xJc_j)^2$$

$$F_4(\text{ligament})= \underset{j=1,8}{\text{SUM}} (P4xL_1)^2$$

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 P1,P2,P3 and P4 are, at this time, constant coefficients de ermining 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 P1,P2 and P3 were set respectively to be the inverse of the maximum muscle pull, joint shear and joint compression. In this way the muscle and joint stresses will tend to achieve their maximum value for equal values of Fi.

At this point, only P4 needs to be determined. This can be done by imposing the requirement that the optimized model output show reasonable agreement with the experimental results.

We would like to note that P4 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.

BIOLOGICAL LIMITS - Equality and inequality constraints -

EQUALITY CONSTRAINTS :The solution of our 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, namely:

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, during 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:

$$\underset{k=1\,to\,N}{\mathrm{SUM}} (A_{ijk} \times K_k) + E_{ij} + L_{ij} + J_{ij} = 0$$

Where:

Aijk = 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.)
Kk = stress in muscle group k
Eij = force component i at joint j due to an external load
Lij = component i at joint j due to the reaction of the ligaments at joint j
Jij = force component i of the reaction at joint j due to the disc and the facet

The Aijk terms can be determined from anatomical descriptions of the cervical spine and x-rays (Appendix A). The values for Eij can be obtained from a precise description of the task to be performed.

The mathematical objective is to determine the muscular action (Kk), the ligament tensions (Lij), and the joint reactions (Jij) due to an external load Eij with the motion of the neck restricted to the sagittal plane.

INEQUALITY CONSTRAINTS

We must now examine the inequality constraints. With the convention that positive stress represents tension, the inequality constraints may be expressed as follows:

$$K_k \geq 0 \qquad \text{and} \qquad L_k \geq 0$$

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 = F1 + F2 + F3 + F4) in such a way that the equality and inequality constraints described above are satisfied.

DETERMINATION OF THE COEFFICIENTS P1,P2 and P3.

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 P1,P2 and P3, one chooses the best available estimate for the maximum values of the muscle pull, joint shear and compression. The initial approximations taken are:

$$P1 = 1. \qquad P2 = .01 \qquad P3 = .01$$

More accurate estimates of these values will be determined by trial and error using experimental data. Since the ratio of the coefficient is what really determines the optimal solution of the minimization problem, the coefficients were normalized in such a way that P1=1.

The significance attached to the ligament weighting factor P4 is different. The parameter P4 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 P4.

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 P4 to a much lower value. The determination of P4 is done by tuning the calculated muscle response to the measured EMG response.

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 would be desirable if some simplification could be found. Such simplification exists because we have restricted the motion to lie in the sagittal plane. As such each muscle will not behave independently. In other words, the muscles will be expected to fire in groups. Determining the minimum number of independent muscular groups is therefore an important step since it will result in a drastic reduction in the computational burden necessary for solving this problem.

The determination of which muscle belongs to what group is a trade-off between reducing the number of variables to a manageable level without sacrificing the freedom necessary to execute the task. First of all we must define the properties that a muscular group should or should not enjoy.

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

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 : MUSCLE GROUPING

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 P2,P3 and P4.

TUNING OF THE PARAMETER P4 Increasing the value of P4 has the effect of forcing the model to rely on muscular action to support the load. The values of P2 and P3 are set to the initial value of .01 as mentioned earlier. P4 is initially set to zero and gradually increased until the muscles reach their maximum for the maximum extensor effort obtained in the experimental investigation (Section 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], we will assume that the maximum voluntary muscular effort performed by our volunteers will not exceed 2/3 of this ultimate limit.

With this assumption the following values of P4 for the flexed, upright and extended neck postures are respectively 0.04, 0.09 and 0.10 .

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

TUNING OF THE PARAMETER P2 The value of the parameter P2 was increased and decreased around its initial guess to determine the consequences on the load distribution in the neck. Changes in P2 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 initially the four parameters P1 to P4 were the only degrees of freedom available in our 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 P2 for all eight intervertebral joints resulted in poor correlation between the simulation and experiments. With a value of P2 = 0.01, the onset of the calculated splenius capitis activity was in the range of 0 to 1kg . By increasing P2, little change in the model response was observed. By decreasing P2, 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).

A separate weighting coefficient was assigned to the upper two cervical joints (occipital-C1-C2). By augmenting the value of this parameter, we noted a significant increase in the extensor effort required to recruit both the splenius capitis and the omohyoid. The response of the model remained insensitive to changes of the P2 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 essentially zero shear component. The entire cervical system will react strongly to the presence of any shear in this area.

The best values for the parameter P2 are 8.0 for the upper two cervical joints and 0.03 for the lower 6 cervical joints.

TUNING OF THE PARAMETER P3 The parameter P3 controls essentially the compression at the joints. By increasing P3 above the initial value of 0.01 we noted a shifting 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 P3 also resulted in the elimination of the omohyoid's contribution.

On the other hand, by decreasing the value of P3 we noted little change in the response of the model. This suggests that the lower values of P3 may be correct, because they indicate a higher compressive strength. A value of 0.01 is used in all subsequent simulation since it represented the most conservative estimate.

SIMULATION OF EXPERIMENTAL TASKS

Using the weighting coefficients selected above, the model was used to simulate the five tasks that the volunteers performed in the experimental investigation (section 2). The line of action of the resistive force of the restraining strap was 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.

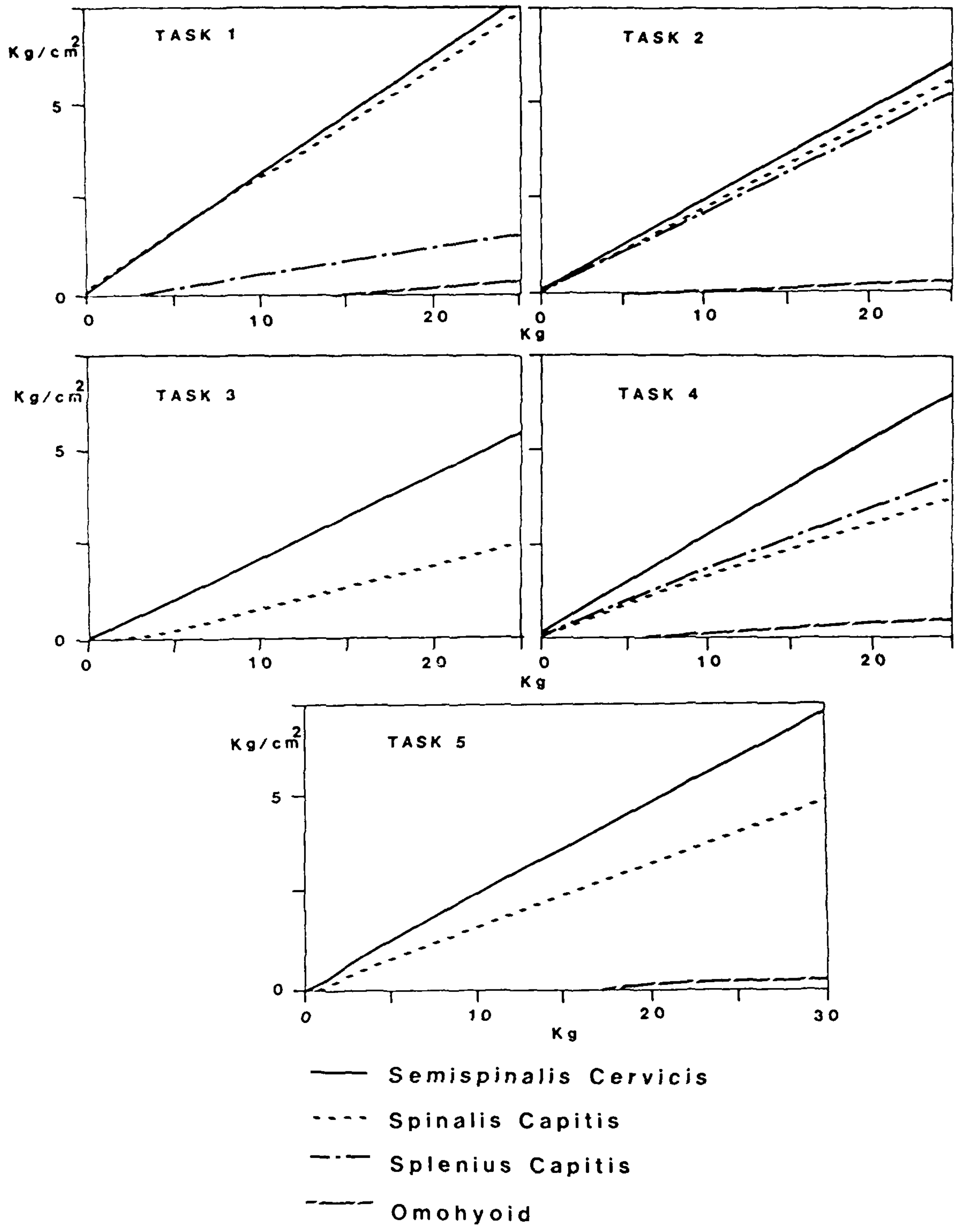


Figure 3.2 Simulation results for neck subjected to loading resulting from subject executing task 1-5

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.

DISCUSSION

The five tasks have been simulated with the same values of P2 and P3. The parameter P4 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.

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 were 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 possible to X-ray volunteers for such determination. Also the restraining strap represents an additional constraint on the subject's attitude while performing the tests.

We also note that the calculated muscular pattern results from an optimization procedure, and that the procedure is only as good as the objective function. The variations in the muscular response from the various volunteers and even the variations observed between tests on the same volunteer cannot be precisely accounted for in such deterministic modelling.

THE RELATION BETWEEN THE CALCULATED RESULTS AND THE 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 we attribute to some small positional shift of the restraining strap. There is also low level activity in the scalenes which we believe is 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 we cannot conclude that we have truly represented physiological behavior we have closely approximated the system of loading in the spine. This approximation appears to be identical in all major respects to the physiological system of loading of the lumbar spine.

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. We must also consider changes in the subjects posture. First there is the case in which the moments generated by the acceleration load is supported only by the musculature (MUSCLE STRATEGY). This will serve as a worst case analysis of how much acceleration the neck can support. Then we will introduce the ligaments as significant moment supporting structures (LIGAMENT STRATEGY) to obtain a more realistic analysis of the cervical spine load bearing capacity.

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 P1 to as large a value as possible without altering the muscle firing strategy by an unreasonable amount. A value of P4 = 1.0 was used.

Results were calculated for the normal upright (or neutral), flexed and extended postures.

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

The relative arrangement of acceleration vectors with respect to the cervical spine was changed as indicated in Fig 4.1. The maximum acceleration supportable for each direction of the acceleration vector was calculated in accordance with the above procedure. The results are shown in Fig. 4.2 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.

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.

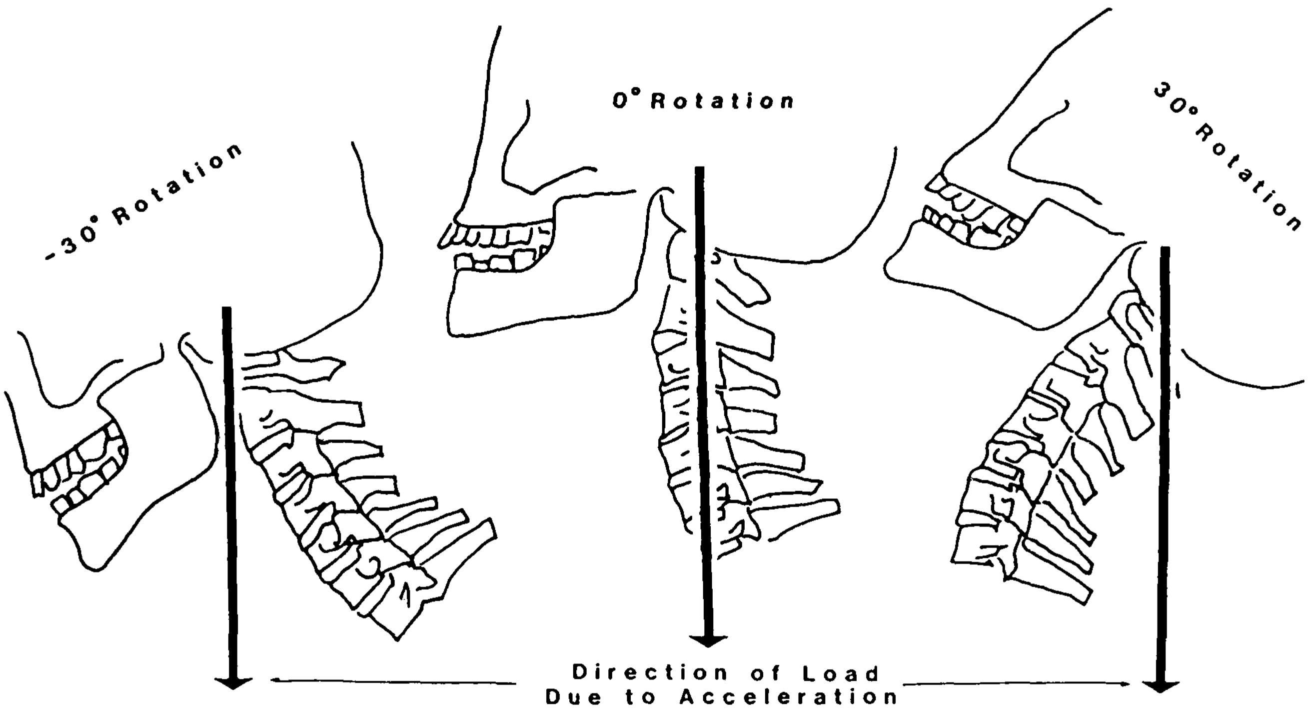


Fig - 4.1 Loading configuration for simulation of high acceleration in neutral spinal geometry.

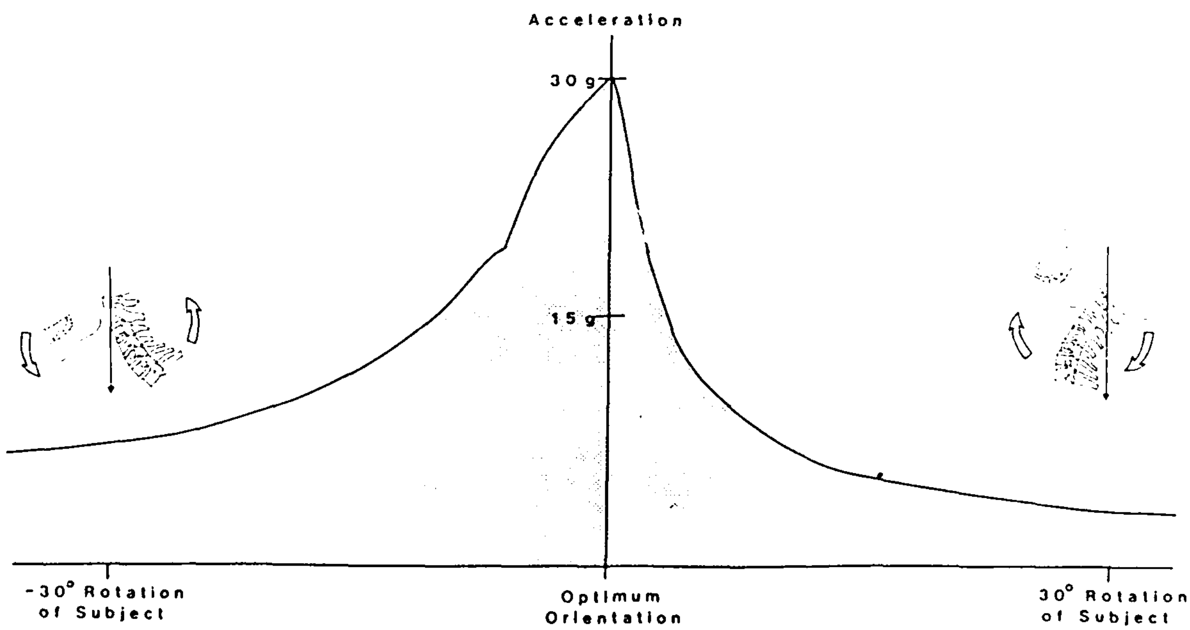


Fig - 4.2 Simulation results for neutral spinal geometry.

FLEXED GEOMETRY This simulation considers the spine to be in a flexed geometry. As was done in the previous case, the acceleration was 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 was varied (Fig. 4.3). Values of maximum acceleration which the muscles could support were determined. These results are presented in Fig. 4.4 as a plot of the maximum acceleration that the muscle can support 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.

EXTENDED GEOMETRY Finally simulations were performed for the neck in the extended spinal geometry. As was done in the previous two cases, the acceleration was increased until one of the cervical muscles 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 ot the spine was varied (Fig. 4.5). Values of maximum acceleration were determined and the results are presented in Fig. 4.6.

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 throughout the spine. The important feature is that the maximum acceleration dropped from 30g for the neutral position to 15g for the flexed position.

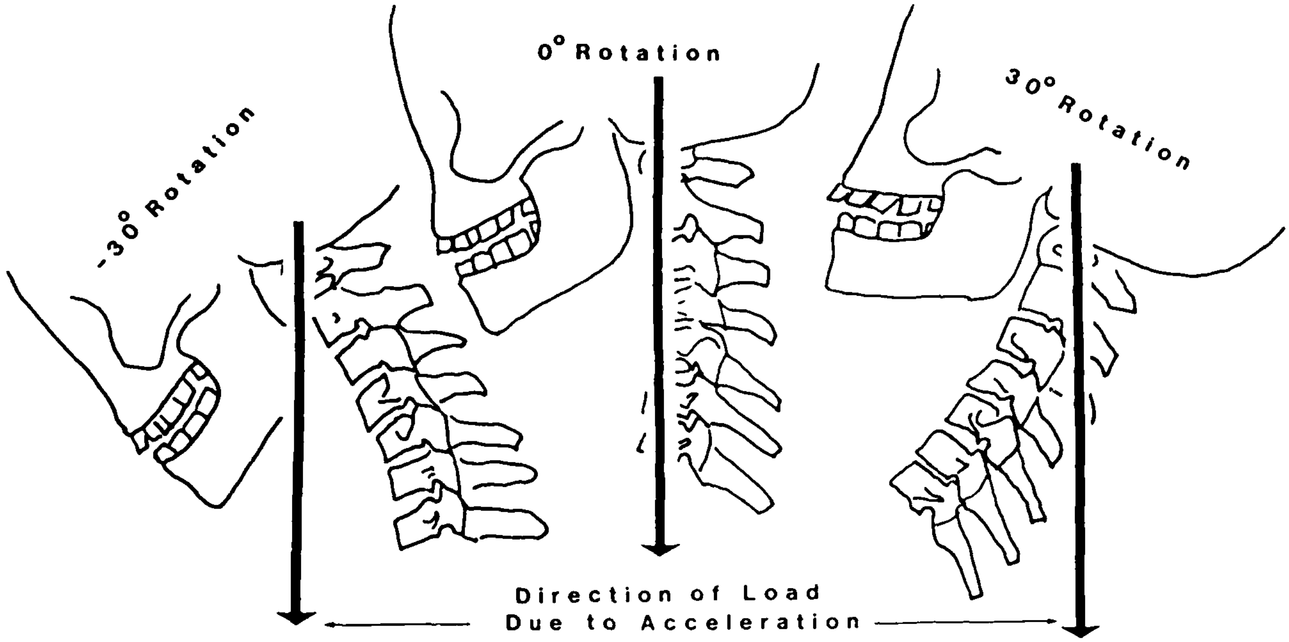


Fig - 4.3 Loading configuration for simulation of high acceleration in flexed spinal geometry.

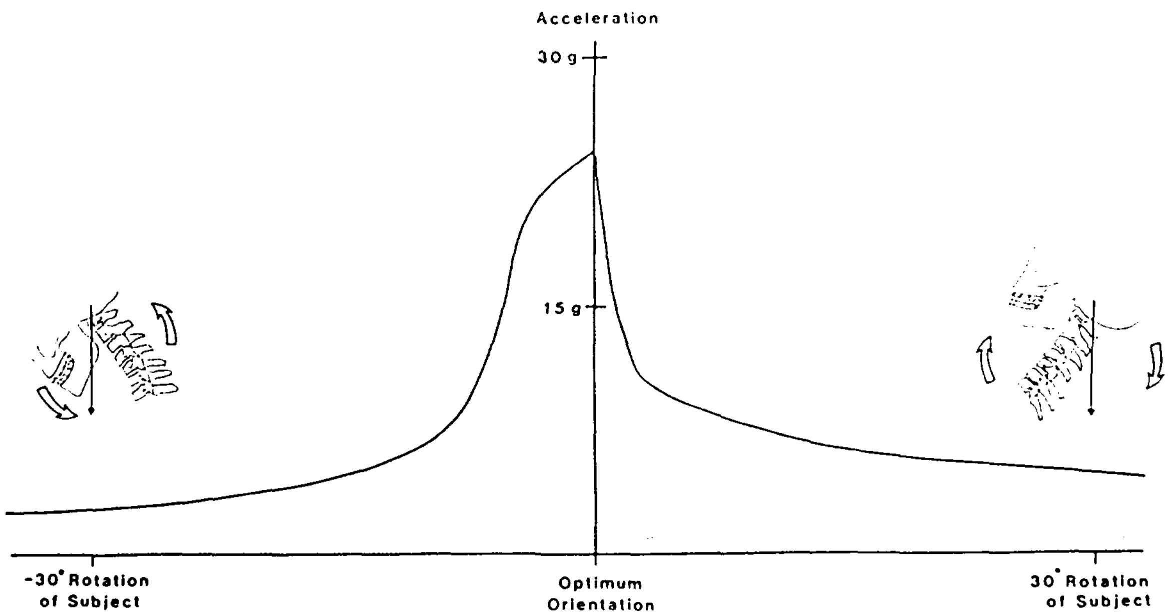


Fig - 4.4 Simulation results for flexed spinal geometry.

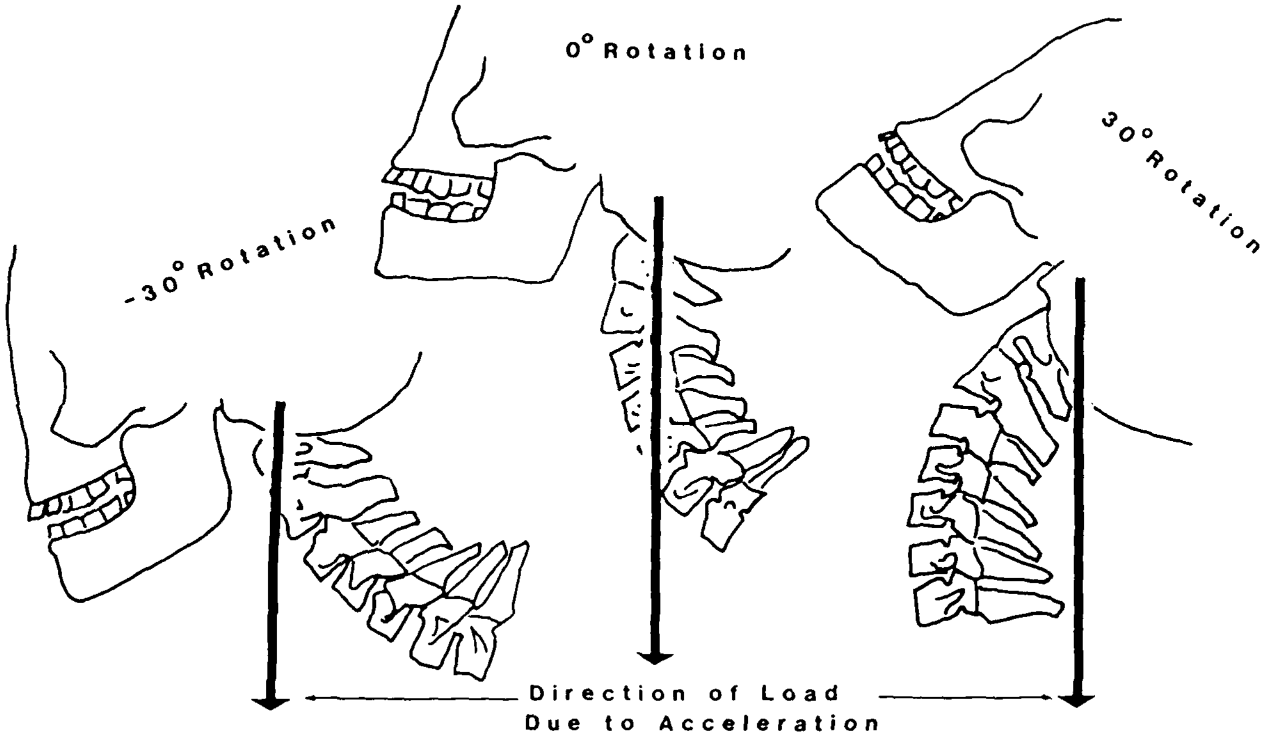


Fig - 4.5 Loading configuration for simulation of high acceleration in extended spinal geometry.

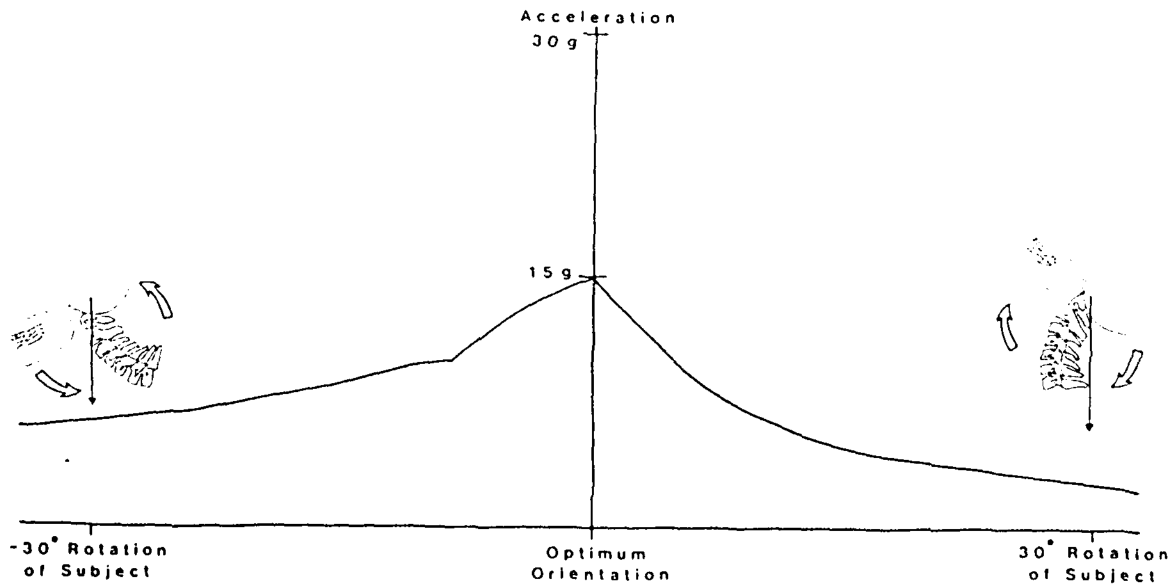


Fig - 4.6 Simulation results for extended spinal geometry.

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.

For each of these two acceleration directions we consided two neck postures, a normal upright posture and a fully flexed posture. Hence four cases must be analyzed.

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 we do not know, but we can estimate the maximum voluntary effort by using simulation results of the experimental tasks presented earlier. We recall that 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 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 were not true, then some of the structures will 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 section 2 & 3 we obtain the following estimated values of voluntary limit in the normal upright posture.

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

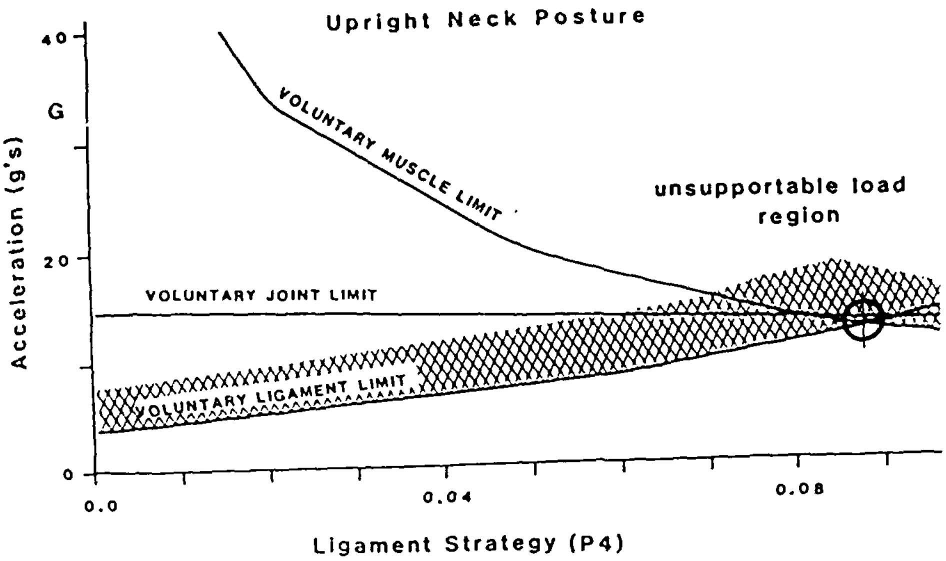
For the fully flexed posture we obtain the following voluntary limits.

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

Recall that the objective function has four parameters Pl to P4. The parameter which remains to be determined in this case is the ligament weighting parameter P4. 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 P4. This has been done for two neck postures and two acceleration directions. (see Fig. 4.7 and Fig 4.8). These figures show the area in which a solution may be found.

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. 4.7). When the acceleration direction is acting at 90 degrees of the compression direction we calculate a maximum supportable value of 13 g's for both postures (Fig. 4.8). The values of P4 are different, namely .0875 and .028 for the upright and flexed postures respectively.

We note 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 remarkable result may be attributed to the fact that the cervical spine is indeed optimized for any task over its full range of motion. From a mathematical standpoint, the solutions for each posture were calculated independently. The similarity of the conclusions is seen as a confirmation that our approach is basically correct. We also believe that Wolff's law is not restricted to the microscopic level of bone structure, but is probably valid in generalized form for the entire macroscopic spinal structure. The proof of this argument is left for future study.



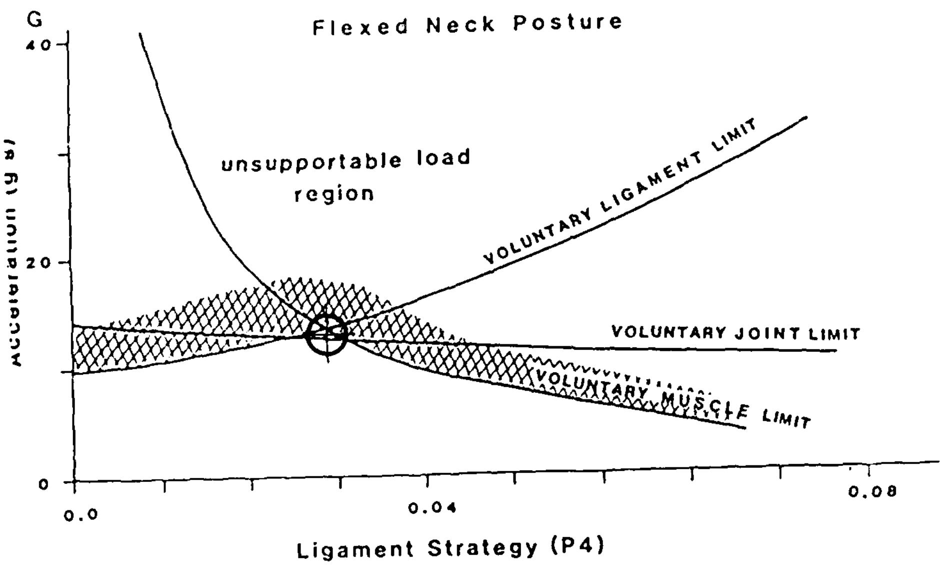
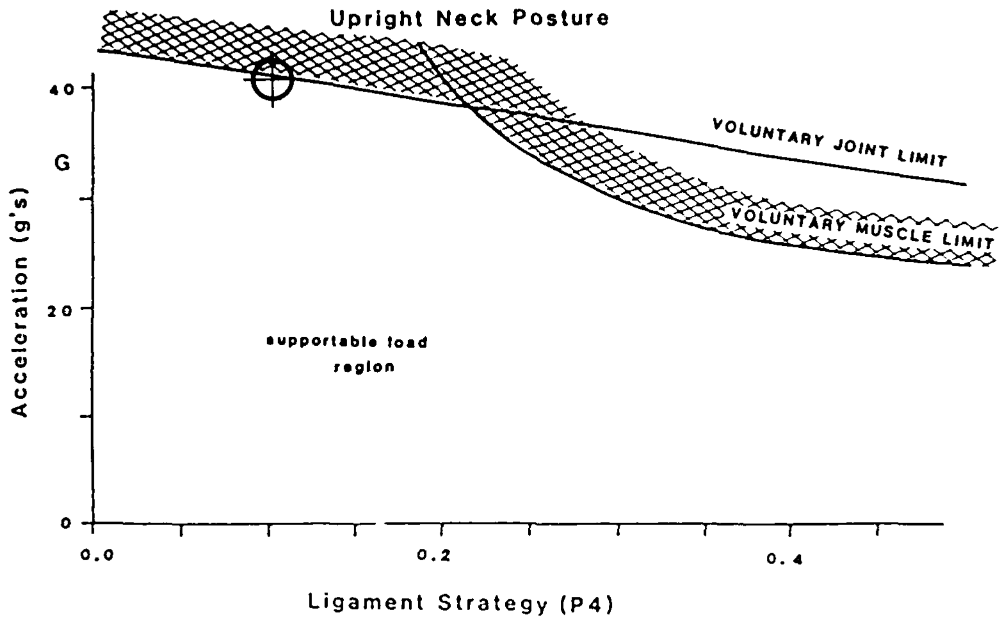


Figure 4.7 Maximum voluntary limits for acceleration loading in the normal upright neck posture and the fully flexed neck posture with load acting perpendicularto axis of spine



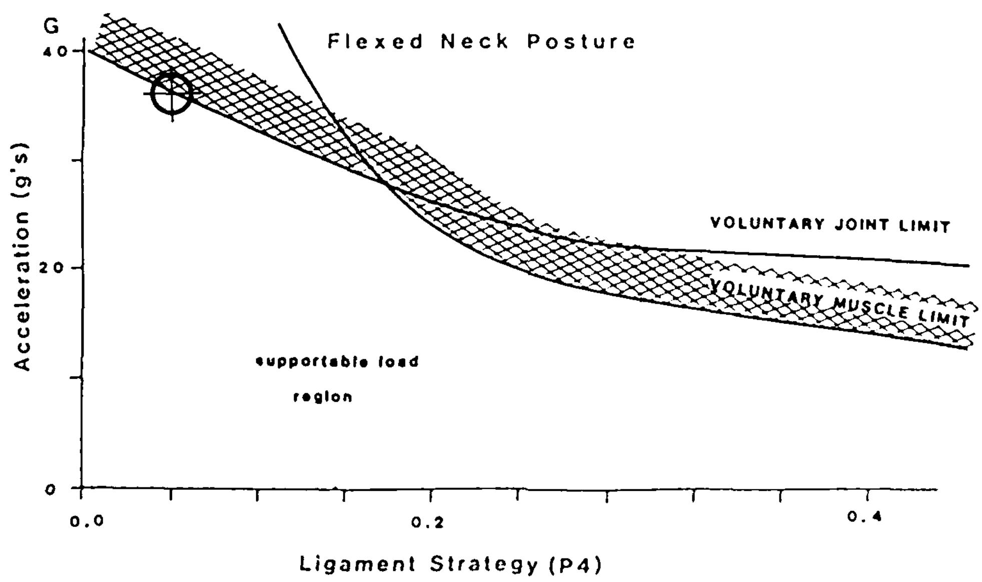


Figure 4.8 Maximum voluntary limits for acceleration loading in the normal upright neck posture and the fully flexed neck posture with acceleration load acting approximately through axis of the spine

DISCUSSION and FUTURE WORK

The results of these simulations can be summarized as follows:

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

> a)- The fact that the O/C1/C2 joints cannot tolerate shear;
> b)- The relative alignment of the spinal lordosis to the acceleration vector.

Any ejection configuration which does not respect these two conditions will lead to premature saturation of the muscular system and loss of control. It is clear that the muscle system is crucial for the control of the stress level in the cervical spine. This particular feature is unique to this model and is the reason for not using spring dashpot models such as [6]. However, once the muscles have lost control of the situation, the spring dashpot models should accurately represent the dynamics of the spine.

2) Lowering the level of stress within the structure implies the use of the ligament system, and the corresponding muscular strategy. If the ligament system is not used (eg. the nuchal ligament is relaxed), then the load is entirely balanced by muscular action. In this case, the best configuration of neck orientation and acceleration vector gives us a calculated upper limit of 30 g's. If we consider that the maximum supportable acceleration is 2/3 of this ultimate limit, we obtain a maximum supportable acceleration of 20g's. Note however that if the spine deviates from the optimum posture, then the maximum supportable acceleration drops to about 5 g's.

3) If 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 described in 2) above.

4) If the structural components (muscle,ligament,joint) were allowed to go their limits rather than the assumed 2/3 voluntary limit used in 3 above, the cervical spine could support a much larger acceleration. By extrapolation we would estimate the highest value to be in the range of 50 g's. However, caution must be exercised before accepting such an extrapolation. For safety reasons, the experimental tuning of our model was performed on human volunteers who had full control of their loading conditions. None of these were in the high acceleration range. In principle, our approach could be verified by appropriate animal experiments.

5) The maximum magnitude of supportable acceleration remains constant with changes in neck posture provided the direction of acceleration is appropriate.

6) The resultant of forces passing through the O/Cl/C2 structure must not contain any shear component. This explains why the unrestrained head will tend to orient itself into the direction of acceleration.

7) This work suggests at least three ways to promote safe pilot ejection.

a)- By proper alignment of the spine with the acceleration vector. In particular note the O/Cl/C2 shear condition.

b)- By external stimulation of appropriate muscles before ejection begins.

c)- By external support such as air bags inflated at the time of ejection. Our studies however emphasize caution: the O/Cl/C2 position relative to the acceleration vector is essential. The pilot's central nervous system will always tend to modify the spinal lordosis in order to achieve zero shear at the O/Cl/C2 joints. An air bag may prevent this geometry change. We suggest that the air bag system should have at least two actions: 1) to freeze the pilot in the most advantageous posture given the constraints of the seat, and 2) to offer actual supplementary support for the head. This will be necessary if the seating configuration does not permit optimal spinal geometry.

8) This analysis could be merged with our previous model on the lumbar spine to re-evaluate the ejection problem for the total spine.

9) We understand that there are other aircrafts that subject the pilot to severe lateral forces. A spinal model for this type of forces could be developed along the same principles. Such a model would determine the muscular action necessary to minimize stress within its spinal structure of the pilot and maintain zero shear at the O/Cl/C2 joints. Muscular fatique could be assessed. This type of analysis would lead to compensatory measrures resulting in an improved seat design.

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 was expressed in mathematical terms.

The large number of muscle strands in the cervical spine have been arranged into a number of functional groups, each with its own independent muscular firing density. In order to perform a given task such as balancing an external load, the muscular strategy has been calculated using an objective function which is minimized by optimization techniques. The objective function expresses that the best muscular action will attempt to 1) minimize the stress at the intervertebral joint and b) balance the external load without exceeding the biological limits of material.

Experimental investigations using electromyographic information obtained from volunteers performing isometric tasks were used to tune the model. Model tuning consists of assigning values to a few key parameters in the objective function. The model is considered tuned when the calculated muscular pattern matches the measured experimental integrated EMG pattern.

The model has been tuned primarily to the pattern of EMG response of two of the accessible muscles, the splenius capitis and the semispinalis capitis. The reason for selecting these two muscles is due to the fact that while the capitis exhibited immediate increasing EMG output with increasing extensor effort against resistance, the splenius capitis action was delayed.

The tuned model was then utilized to simulate low acceleration loads. We found that the maximum voluntary effort which can be achieved was dictated by the amount of flexion moment that is generated at the lower cervical spine. This fact is reflected in the model's attempt to recruit the omohyoid muscle when the neck is subjected to very high flexion moments.

Another observation concerns the preferred direction of loading for the intervertebral cervical joints. Any shear at the occipital-atlas-axis joint is undesirable. This indicates that the stress at these levels has a very direct and primary effect on the muscle firing strategy of the cervical spine.

The tuned model was then utilized to simulate high acceleration loads. First, we analyzed the situation that would result if the muscles alone were left to balance the acceleration load. The simulations showed that if the load was acting approximately through the axis of the spine so as to produce high compressions at the intervertebral joint with relatively little shear or moments, then accelerations as high as 20 to 30g could be supported depending on the posture of the neck. As the direction of acceleration deviated from this optimum, the maximum acceleration dropped to levels as low as 5g.

Secondly, we considered the case in which the ligamentous structure has a primary role in balancing the load. Acceleration values up to 40 g's were shown possible without exceeding 2/3 of the best estimate of the ultimate limit of the structural components of the spine.

These results where true for both the normal upright and fully flexed postures. It was also shown that the maximum magnitude of supportable load that the neck can support was the same for both postures.

Caution must be exercised however in interpreting the results calculated for high acceleration loads: the model was validated on human experiments performed at very low acceleration (gravity), and it is this validated model that was used to simulate the spinal response at high acceleration. Proper animal experiments must be designed for validation of our results.

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APPENDIX A - 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:

$$\underset{k=1 \text{to} Nm}{\text{SUM}} (A_{ijk} K_k) + E_{ij} + L_{ij} + J_{ij} = 0$$

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.

DESCRIPTION OF RELEVANT SKELETAL STRUCTURE

In the process of numerically describing the muscles of the neck and the resulting reaction forces that these muscles will produce at the I.V. joints, the vertebrae of the neck (C1-C7) and the occipital bone must be included as points of attachment for the muscles. 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 will 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 structure are described in a local coordinate frame and the skeletal structures are assembled by assigning a location and orientation for 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 positions (upright or neutral, full flexion, and full extension), an assembled skeleton, and individual vertebrae. The points of attachment to the vertebrae were organized in a single array [3 x 17]. Each of the points was 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 was 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.

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 Occipital Bone (see Figure 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.

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

TABLE A.2 ATLAS (C1) (Figure A.1)

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**
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

Table A.3 Axis (C2) (see Figure A.2)

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 were described with 11 points in the local coordinate frame. See Table A.4.

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.4 CERVICAL VERTEBRA (C3 - C7) (Figures A.2)

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

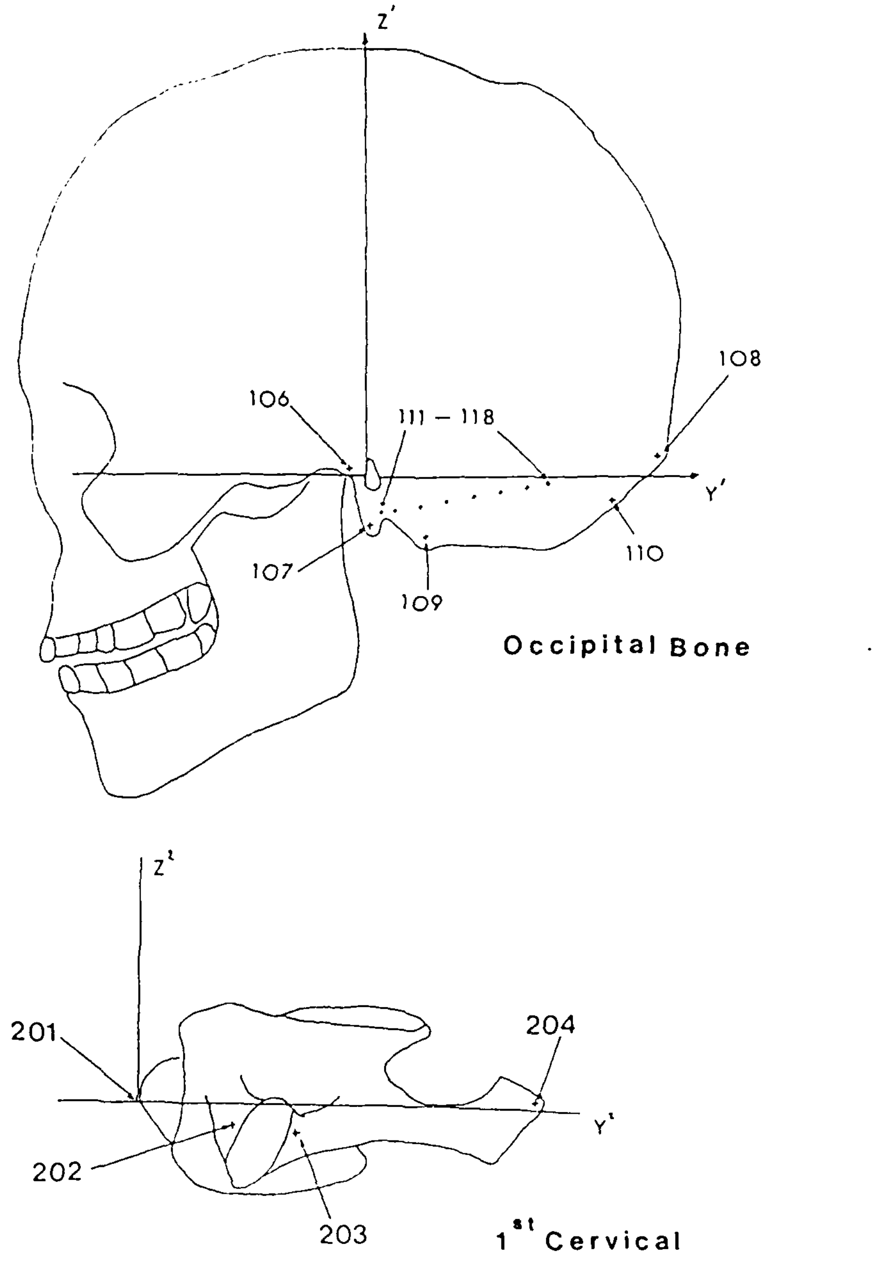


Figure A.1 Points of muscle attachment on the head and atlas

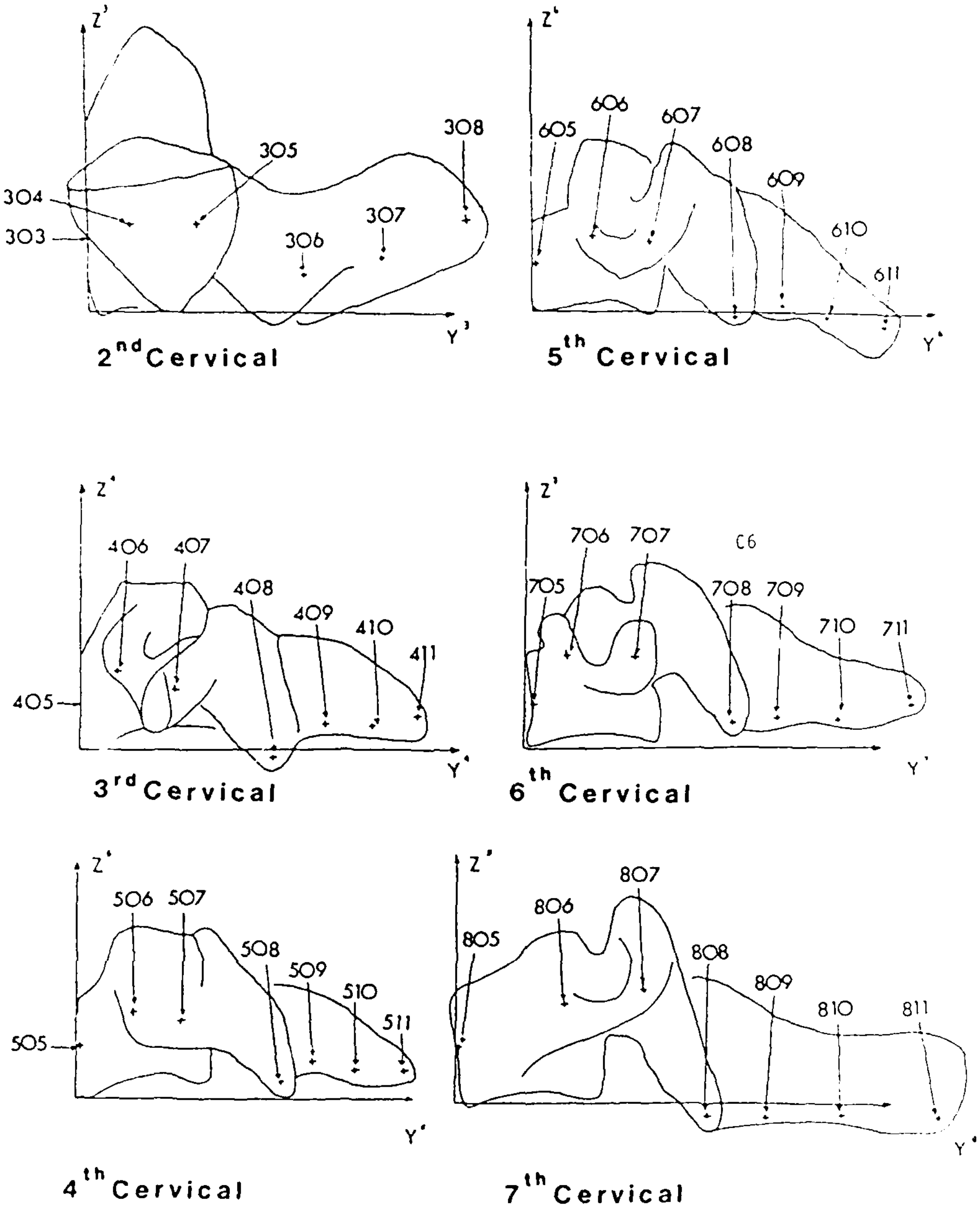


Figure A.2 Points of muscle attachment on the 2^{nd} to 7^{th} cervical vertebra

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

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.5 THORACIC VERTEBRA (T1 - T6) (Figures A.3 and A.4)

The local coordinate frames for the thoracic vertebrae are defined in the same manner as for the cervical vertebrae. The 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).

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.

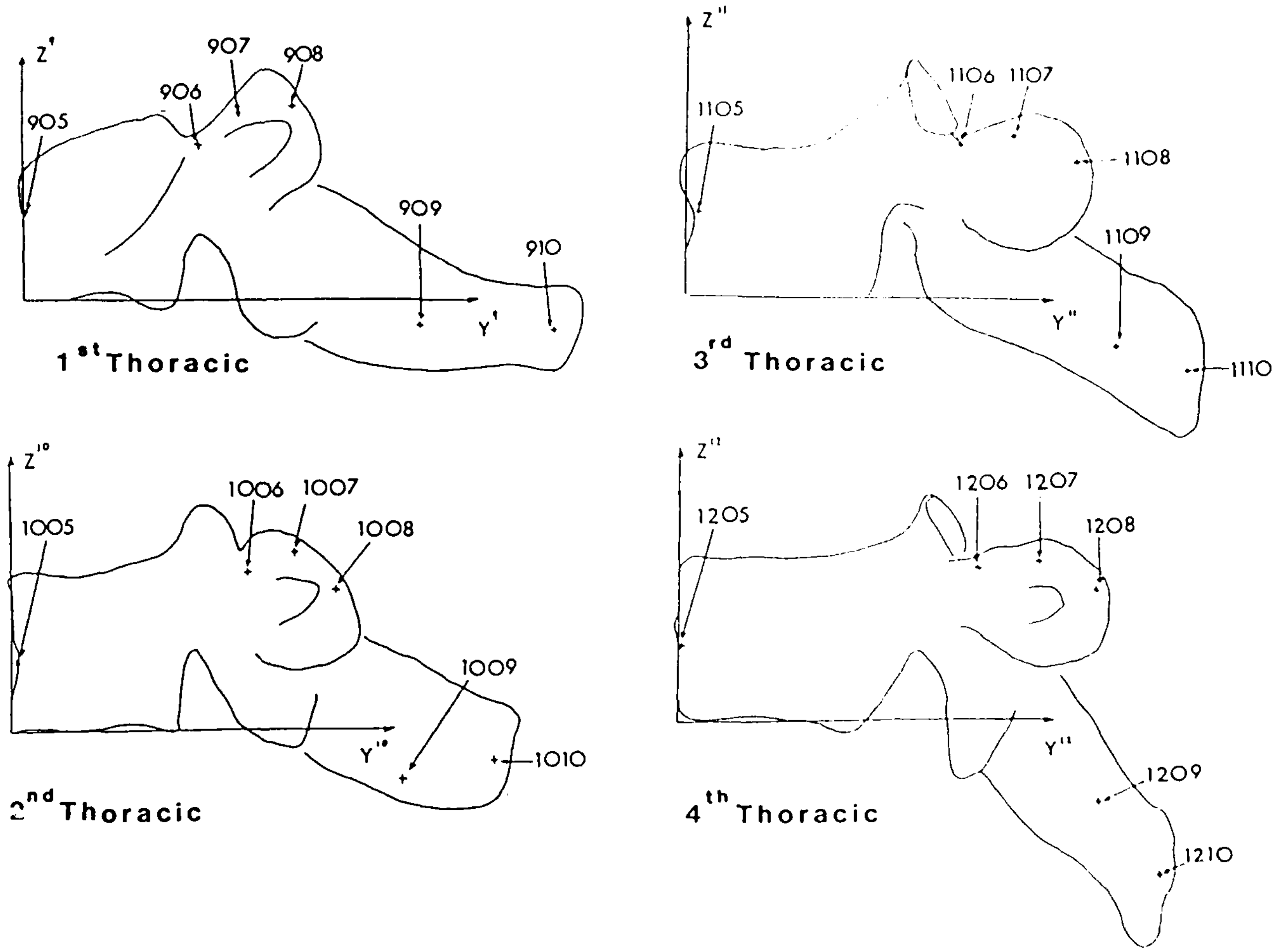


Figure A.3 Points of muscle attachment for the 1st thru 4th thoracic vertebra

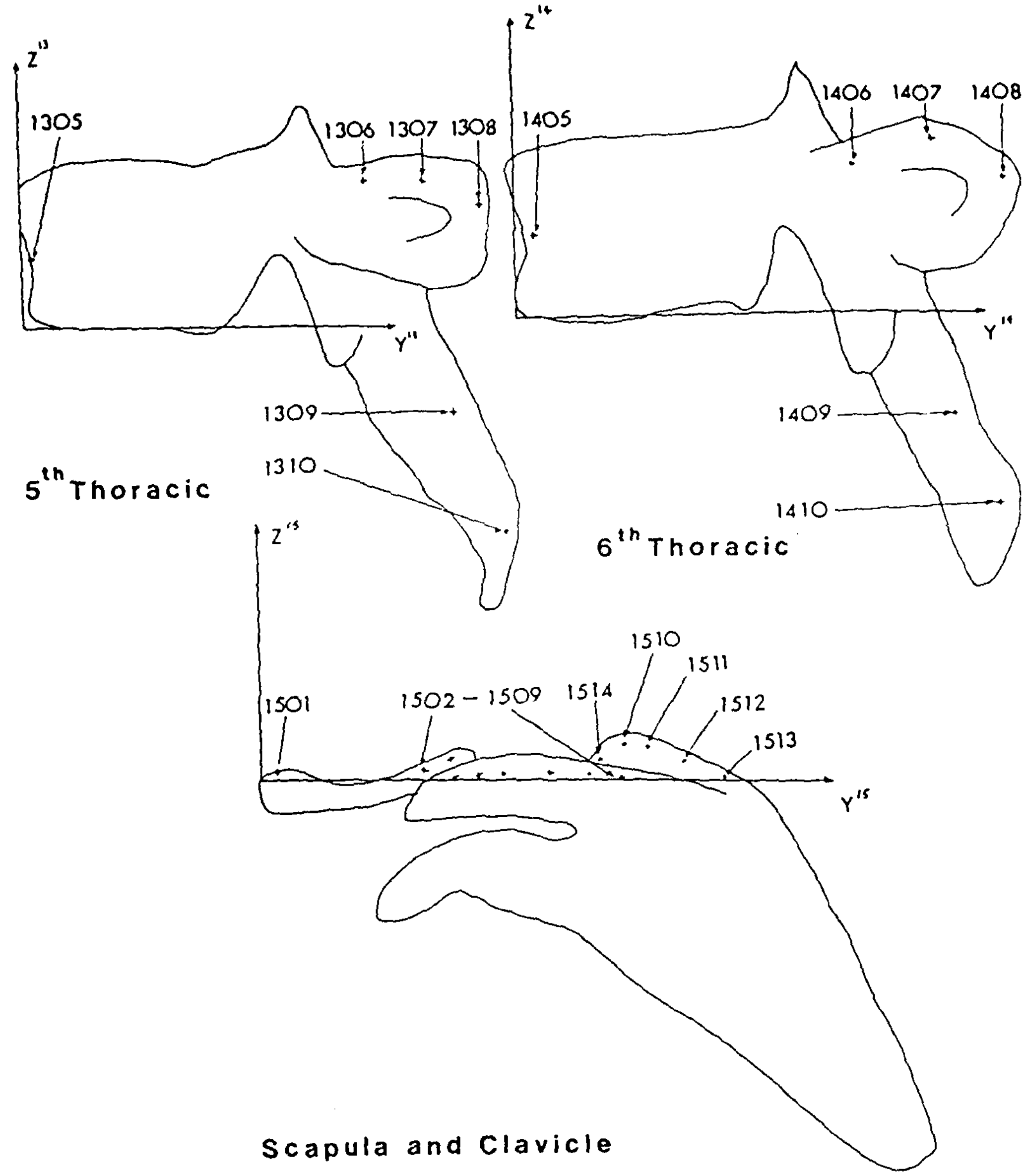


Figure A.4 Points of muscle attachment for the 5th and 6th thoracic vertebra and scapula and clavicle

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-Cl
1512	-	vertebral margin of scapula origin: rhomboid to C7
1513	-	vertebral margin of scapula origin: rhomboid to C6
1514	-	origin: omohyoid

TABLE A.6 CLAVICLE + SCAPULA (see Figure A.4)

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 spine 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).

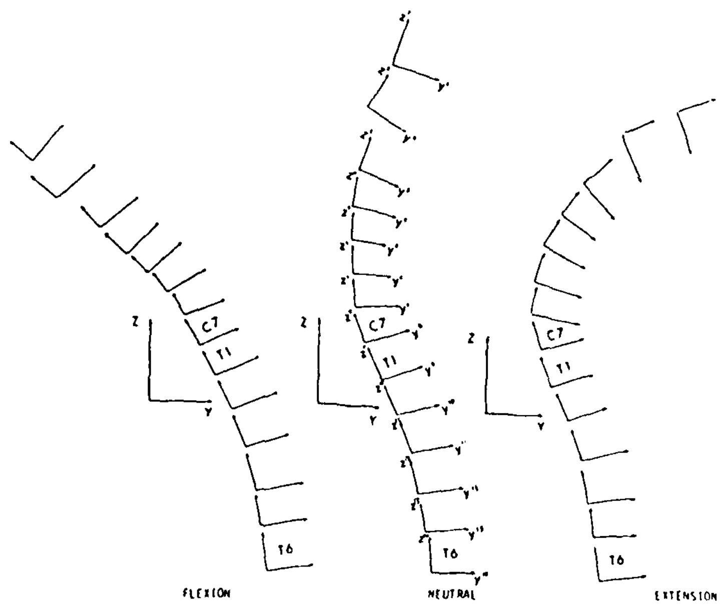


Figure A.5 Representation of three basic postures in global achieved through coordinate transformation

LIGAMENT DESCRIPTION

The neck will be assumed to be in a flexed ,extended, or normal upright posture with the jaw closed at all times. The applied load will be limited to forces acting in a direction (from posterior to anterior) so as to produce flexion moments at the intervertebral joints. No load will be applied to the arms and shoulders. The arms will be assumed to be adequately supported and therefore will 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 (muscles), and the passive structural elements (ligaments, intervertebral disc and articular facets).

When a flexion moment acts on the intervertebral joint, we assume 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.

We have chosen to describe the disc 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 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 was the case for the muscles, but examination of the anatomy suggests that the above idealization is 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, we propose a different approach based on the use of electromyographic information.

While standing erect or while fully flexed, the spine requires very little or no muscular activity to maintain posture [Floyd and Silver,1955]. 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 we can estimate the ligament tension in the fully flexed posture 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 will 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.

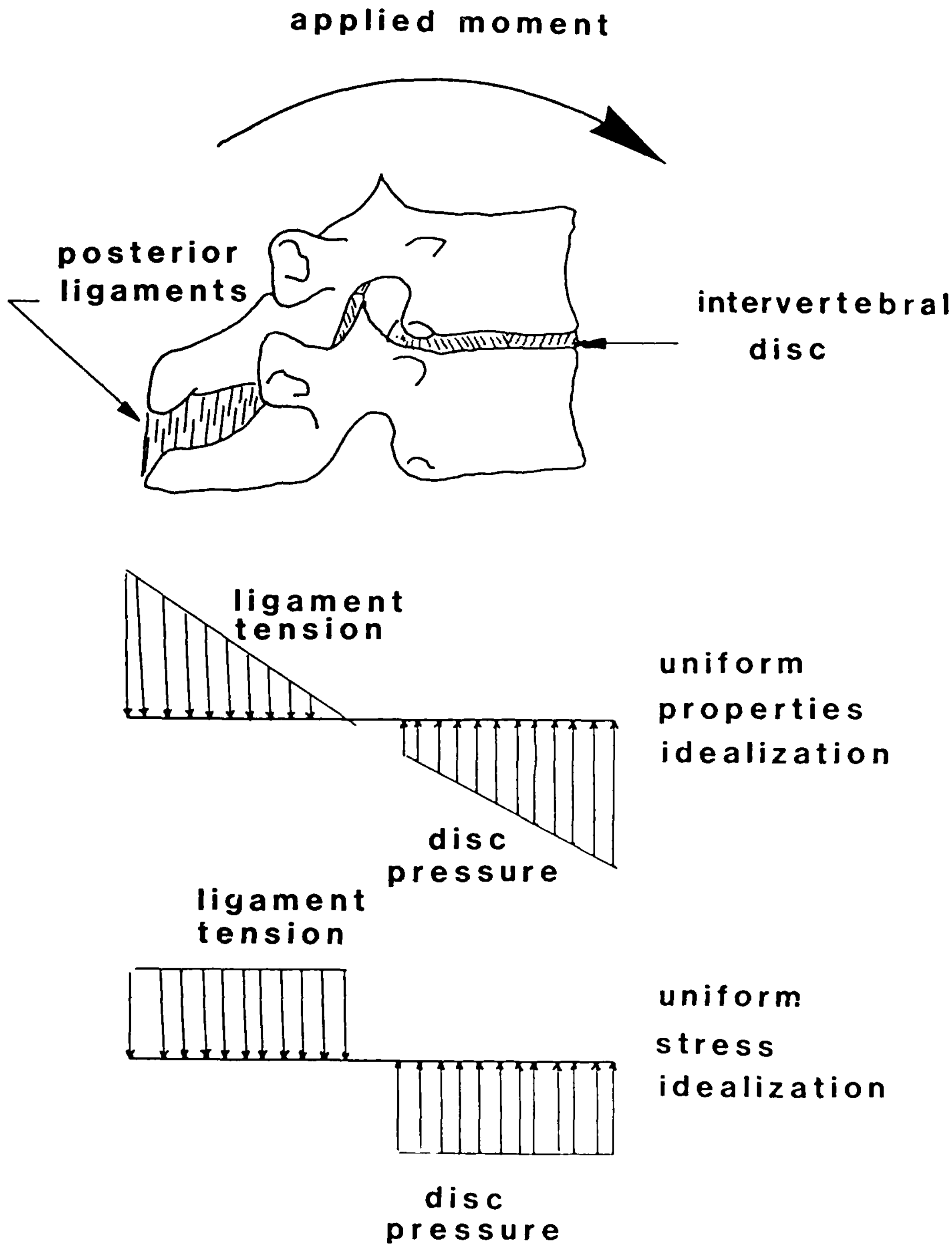


Figure A. 6 Idealization of intervertebral joint and ligaments

JOINT DESCRIPTION

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.

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

$\underline{L}_j$ = unit vector in direction of force of ligament of joint level j

L_j = cross-sectional area of ligament

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

M_k = cross-sectional area of muscle

Shear/unit muscle stress

$A_{1jk} = [\underline{S}_j \cdot \underline{M}_k] M_k$

Compression/unit muscle stress

$A_{2jk} = [\underline{C}_j \cdot \underline{L}_j] L_j$

Moment/unit muscle stress

$A_{3jk} = M_k \cdot \text{mag}\{\underline{r} \times \underline{M}_k\}$

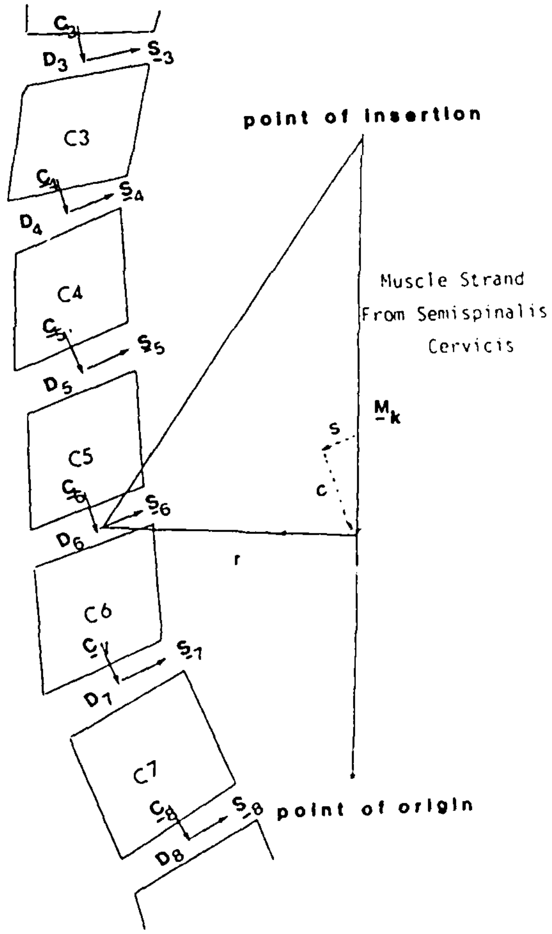


Figure A.7 Muscle and ligament tensions result in shear, compression and moment at the intervertebral joint

MUSCLE DESCRIPTION

The muscles are considered as a collection of 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 will be 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, we can consider the shear (A_{1jk}), the compression (A_{2jk}) and the moment (A_{3jk}) produced at each joint. If we consider 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:

$$A_{1jk} = [\underline{S}_j \cdot \underline{M}_k] M_k$$

$$A_{2jk} = [\underline{C}_j \cdot \underline{M}_k] M_k$$

$$A_{3jk} = M_k \cdot \mathrm{mag}[\underline{r} \times \underline{M}_k]$$

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

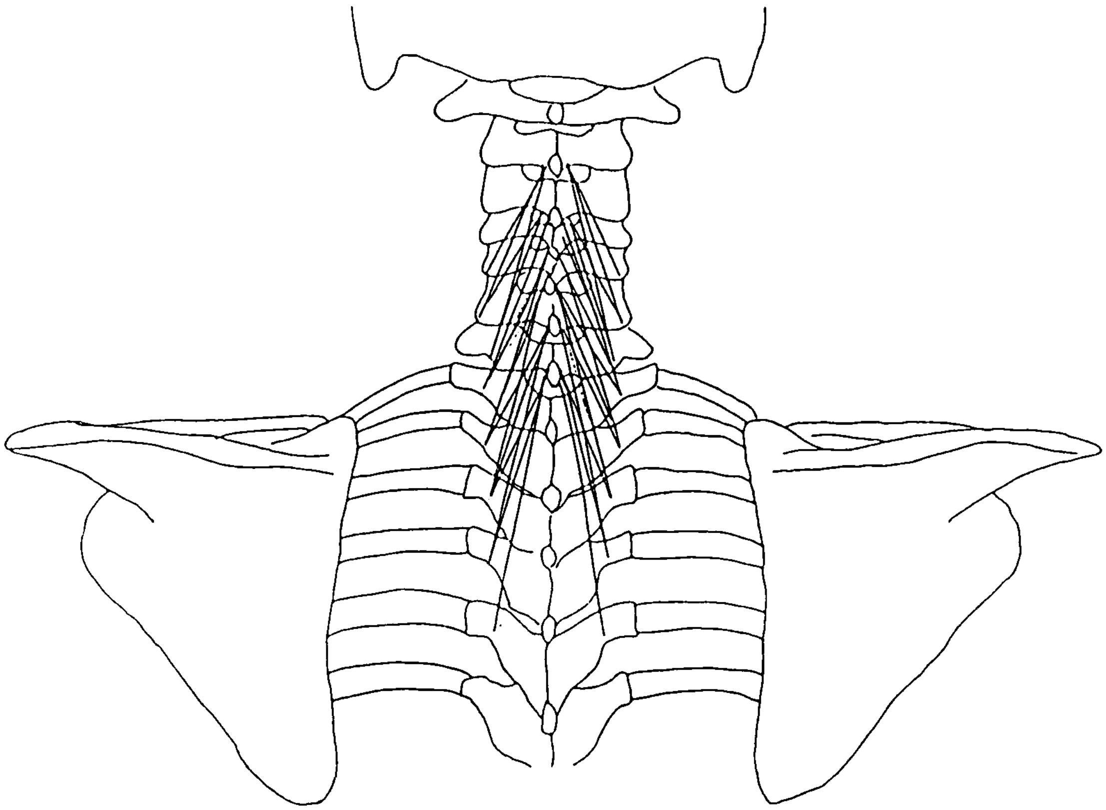
M_k = the cross sectional area of muscle strand k

x = cross product

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.

Multifidus - The multifidus is the deepest of the muscles to be considered in this description of the cervical spine. We note 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).



MUSCLE NO. 1	MULTIFIDUS			
STRAND NO.	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

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 was 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 square was 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 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. We calculated 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.

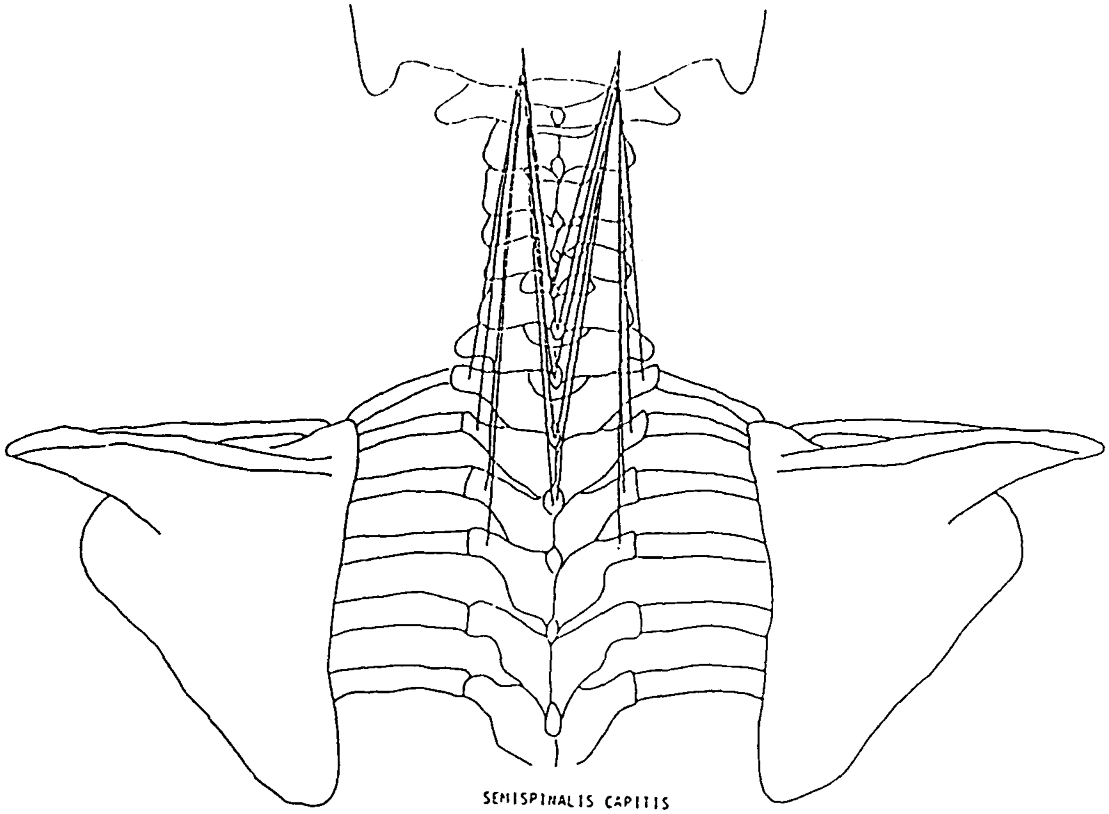
Semispinalis Capitis and Spinalis Capitis - These two muscles are the largest muscles of the neck. Some anatomy books do not differentiate between these two muscles and treat them as a single mass. These muscles cover the multifidus as it arises from the transverse process of T4 to T1 and the articular 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 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.



MUSCLE NO. 2	SEMISPINALIS CAPITAS			
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

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.

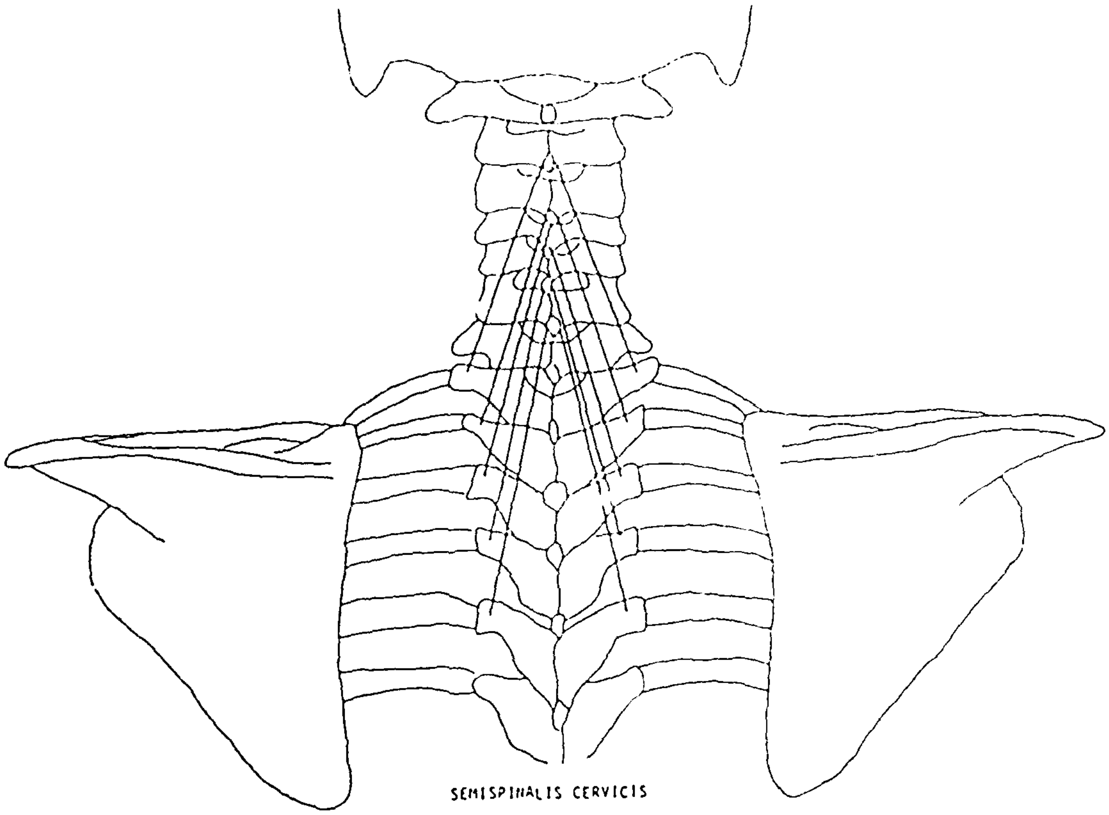
The strands arising from the lower thoracic vertebrae T2-T4 show very little change in length. This is also the case for the multifidus muscle. This can be partly attributed to the fact that the thoracic vertebrae are considered as fixed when in fact they rotate a few degrees in flexion. Another factor is that the thoracic vertebrae were 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. We also 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 we 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).

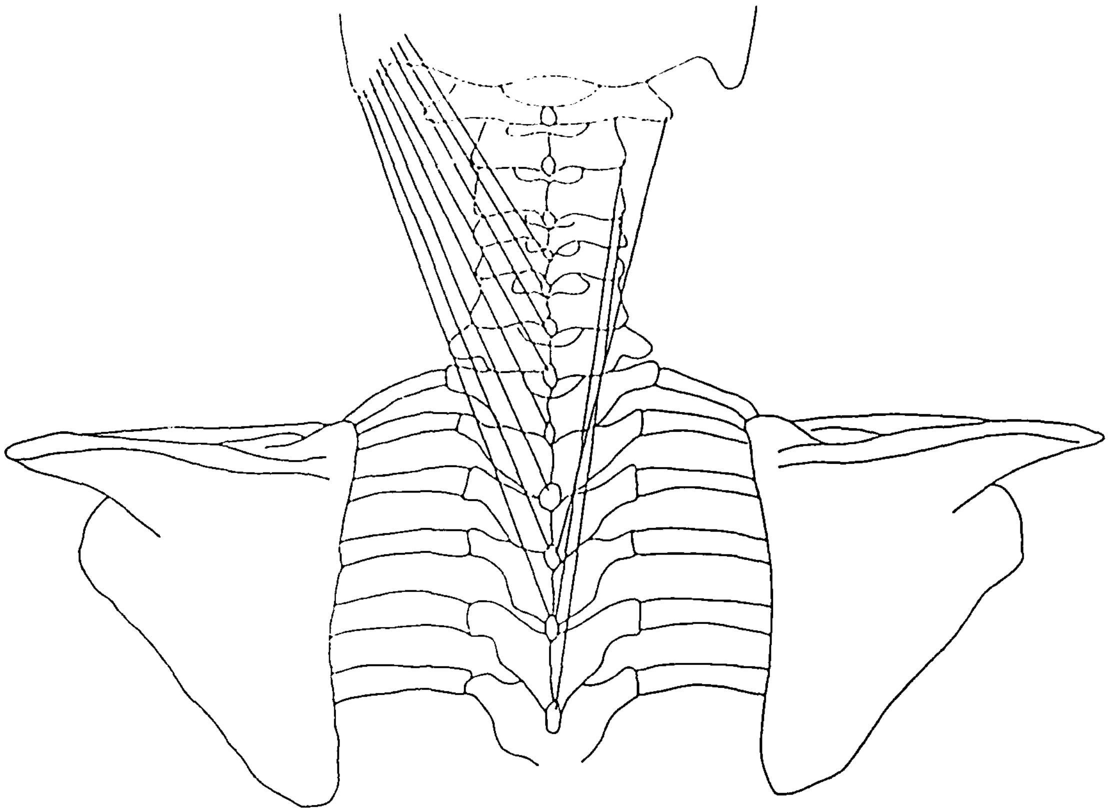
The splenius capitis arises from the ligament nuchae at the level of C4 to C7 and from the spinous processes of T1 to T4; and it inserts into the occipital bone just below the lateral part of the superior nuchae line and into the mastoid process. The muscle is described in our model by 8 strands. The 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.



MUSCLE NO. 3 SEMISPINALIS CERVICI

STRAND NO.	ID NO.	ORIGIN	INSERTION	AREA(CM**2)
37	301	906	307	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 CAPITAS			
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	502	1210	305	0.4
53	503	1310	407	0.4
54	504	1410	507	0.4

Figure A.11 Vector description of the splenius cervicis and capitis

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.

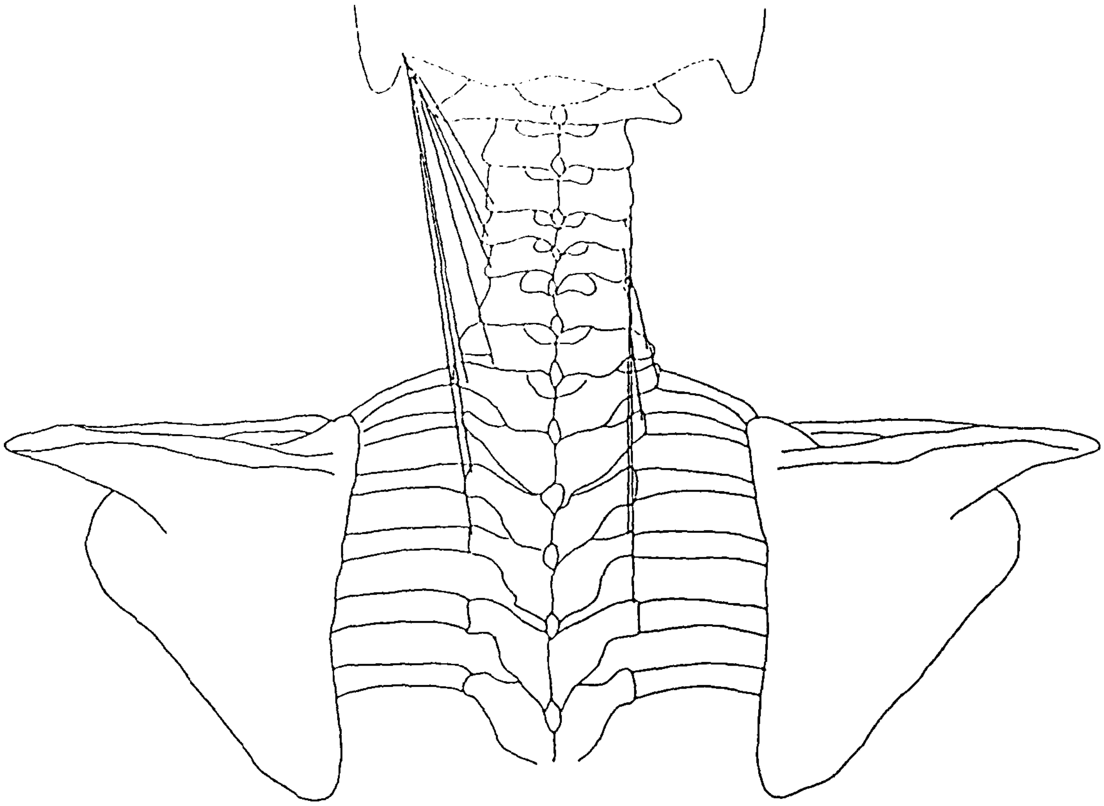
Longissimus Capitis - Cervicis - The longissimus capitis is a slender muscle originating from a)the articular processes of C3 to C7 and b)from the tips of the transverse processes at T1 to T3, and inserting into the occipital bone at the posterior margin of the mastoid process just under the splenius 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.



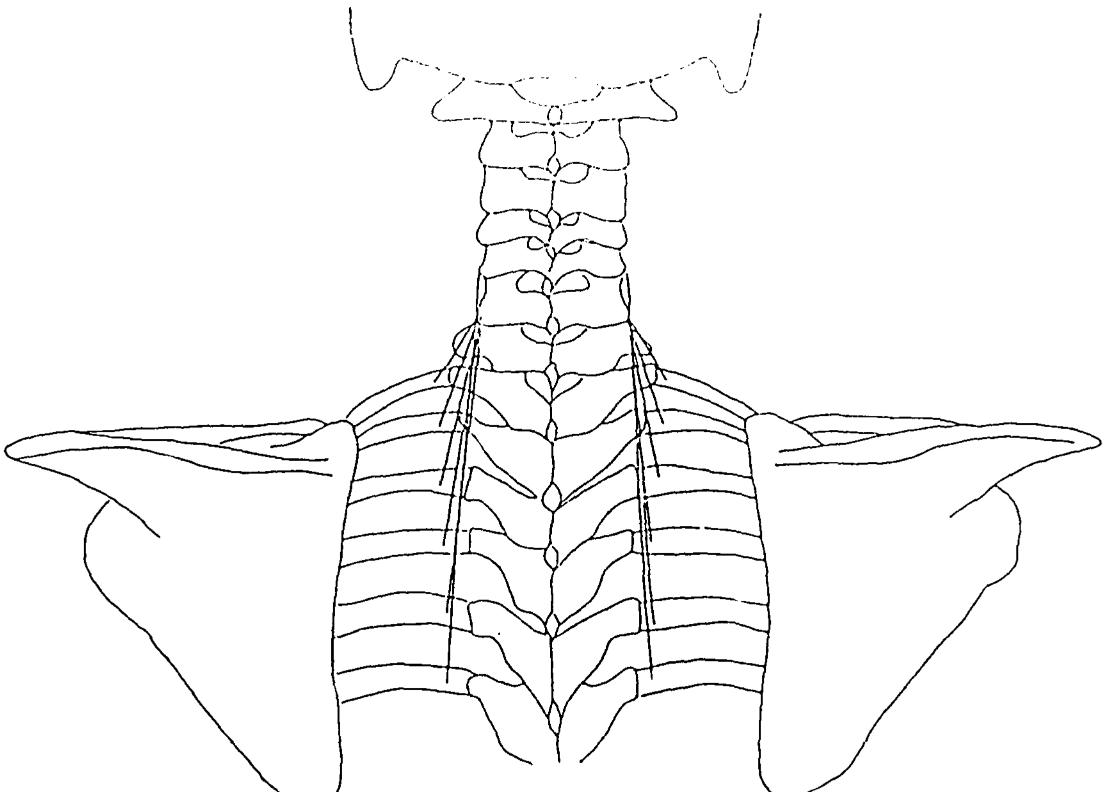
MUSCLE NO. 6 LONGISSIMUS CAPITAS

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 A12 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

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 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.

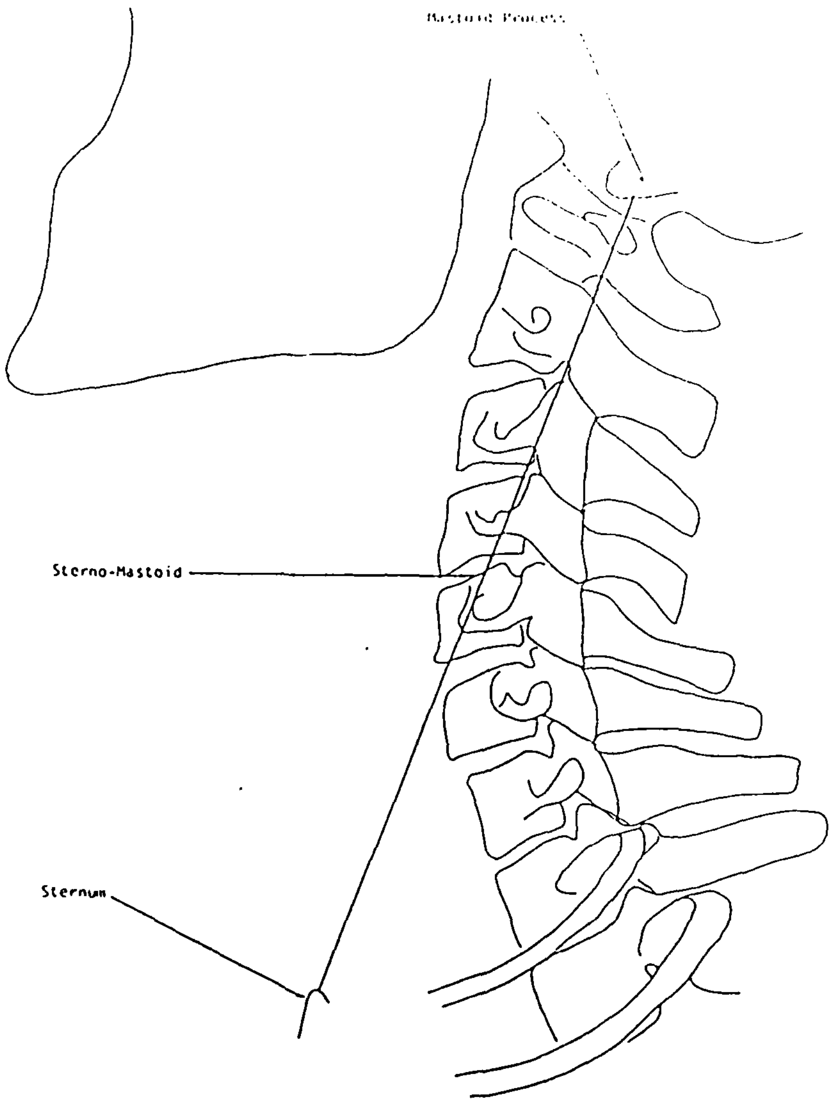
Scalenes - The scalene muscle group is made up of three muscles; scalenus anterior, scalenus medius and scalenus posterior. There is also a scalenus minimum which is considered to be part of the scalenus medius. Each of the scalene muscles is described separately although it is felt that they play the 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 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

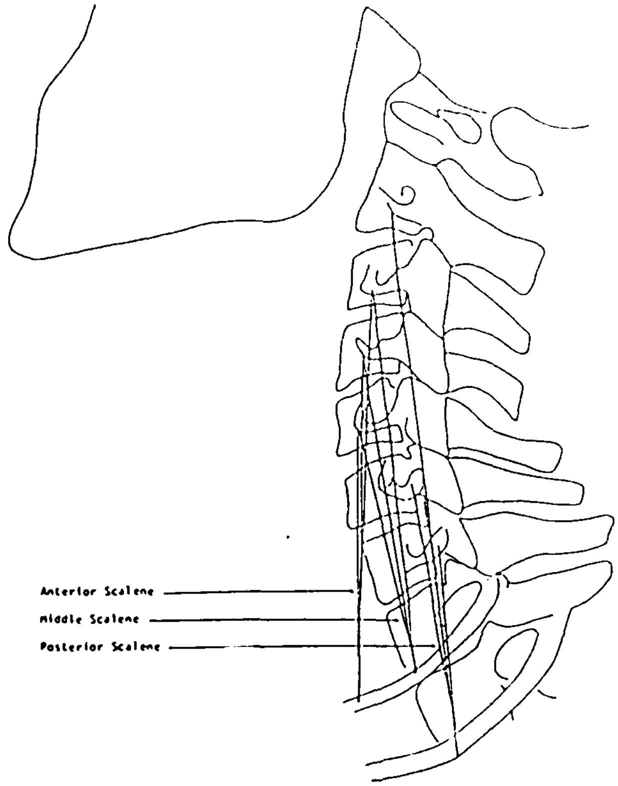
The scalenus posterior, which usually fuses with the scalenus medius, originates from the lateral surface of the second rib and inserts into the posterior part of the transverse processes of C5 to C7. Therefore the muscle is described as three strands originating from a common point of origin and inserting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



MUSCLE NO. 9	STERNO-MASTOID			
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 POST.			
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 MED.			
STRAND NO.	ID NO.	ORIGIN	INSERTION	AREA(CM**2)
78	1101	1602	304	0.4
79	1102	1602	406	0.4
80	1103	1602	506	0.4
81	1104	1602	606	0.4
82	1105	1602	706	0.4
83	1106	1602	806	0.4

MUSCLE NO. 12	SCALENE ANT.			
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

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.

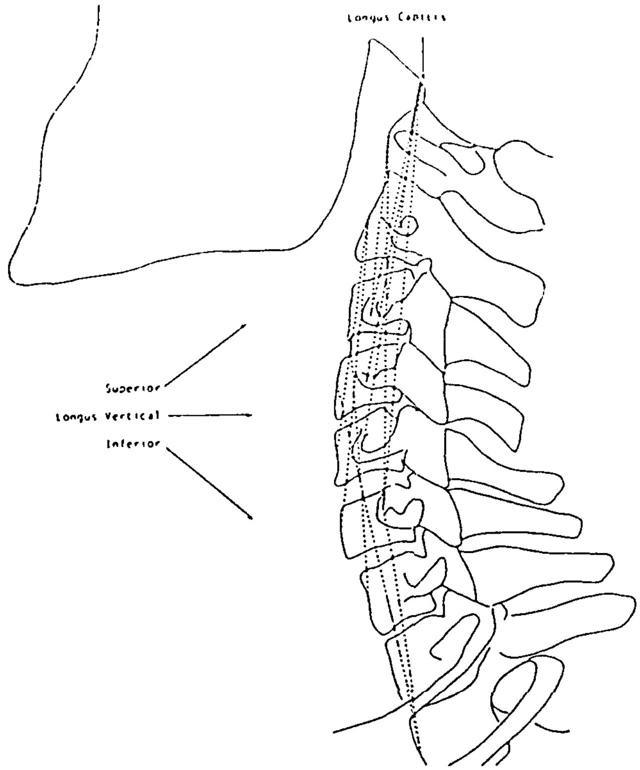
These muscles exhibit some change in length during flexion and extension. The moment produced by these muscles is relatively small. This small moment is due to a small cross-sectional area and to a short level arm.

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 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).



MUSCLE NO. 13 — LONGUS CAPITAS

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

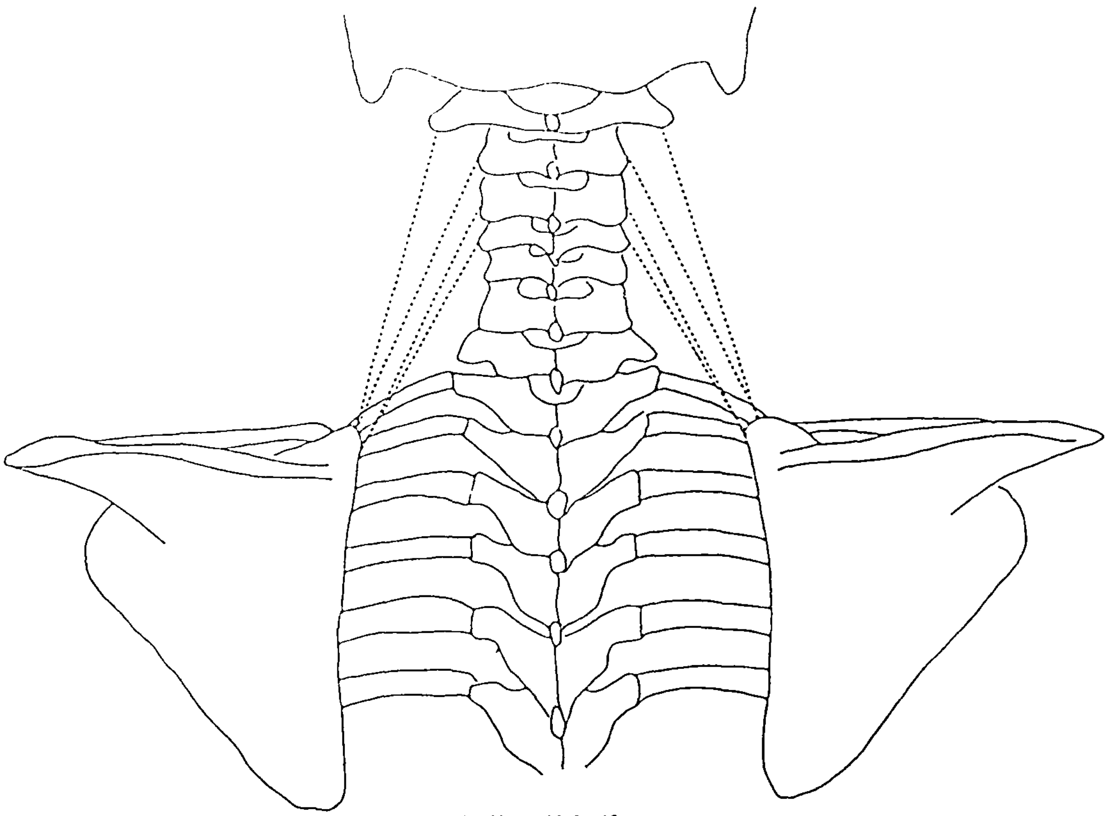
MUSCLE NO. 15 — LONGUS VERTICAL

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

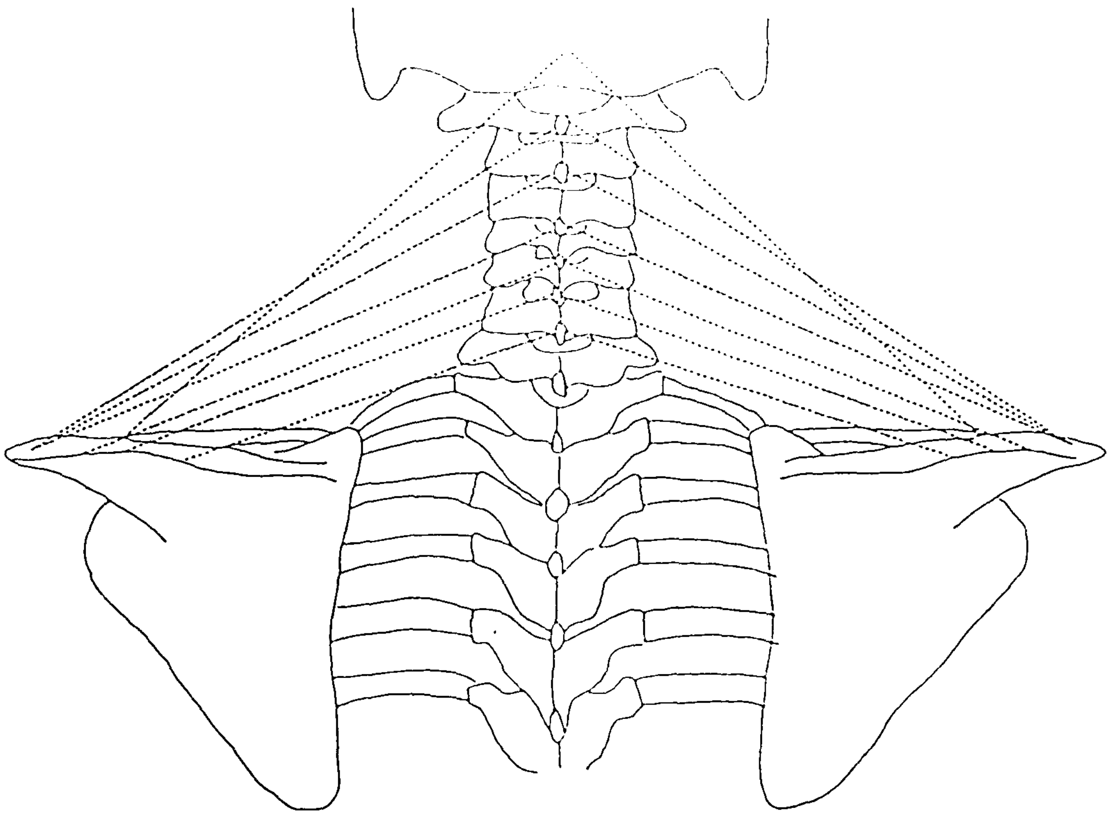
Figure A16 Vector description of the longus capitis and cervicis



LEVATOR - SCAPULAS

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



TRAPEZIUS

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 A18 Vector description of the trapezius muscle

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 we are only concerned with the cervical joints and the rhomboid major inserts only into the thoracic region, we will only consider the rhomboid minor. 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).

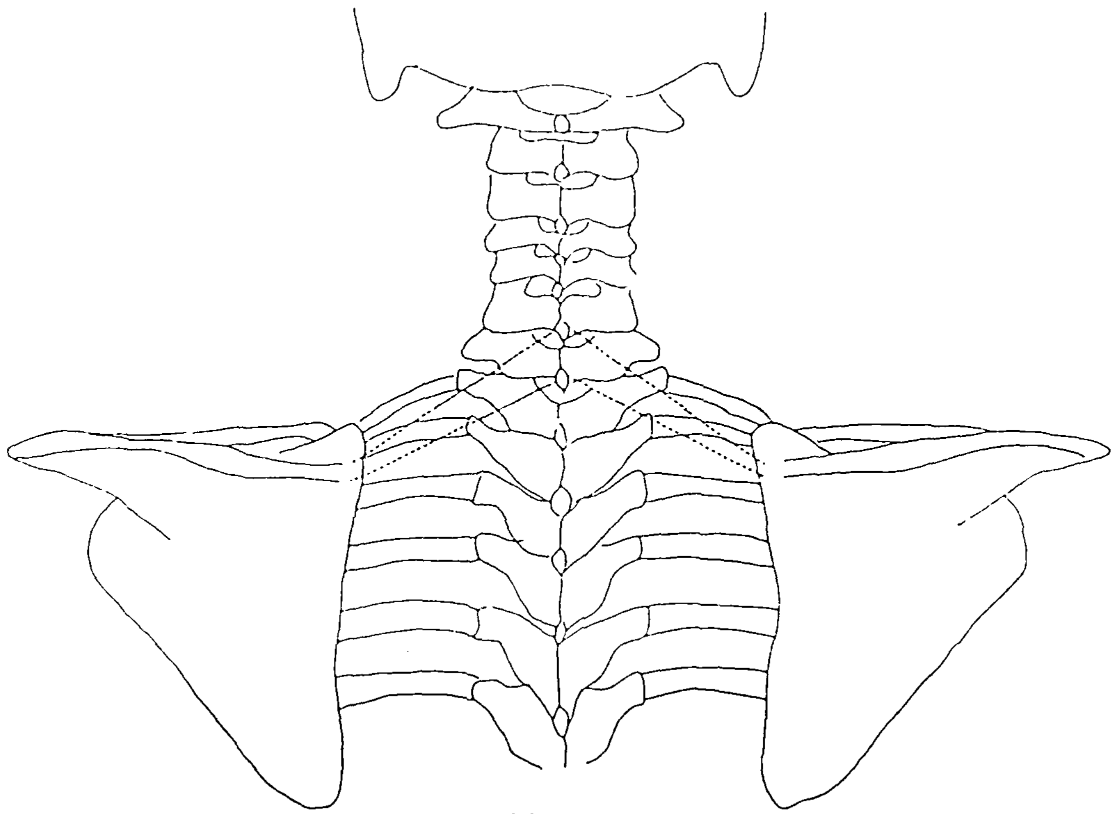
The muscle is modelled with 2 strands originating from 2 points of origin on the scapula and inserting into the tips of the spinous processes of C6 and C7. There is very little change in length during flexion and extension. This muscle produces a large moment due to its large cross sectional area and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

Like the other muscles originating from the scapula, it is assumed that it will 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. We are only concerned with the upper half of this muscle which inserts into C6 and C7. 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

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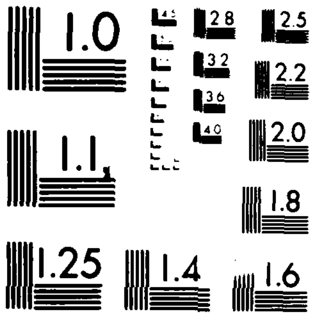

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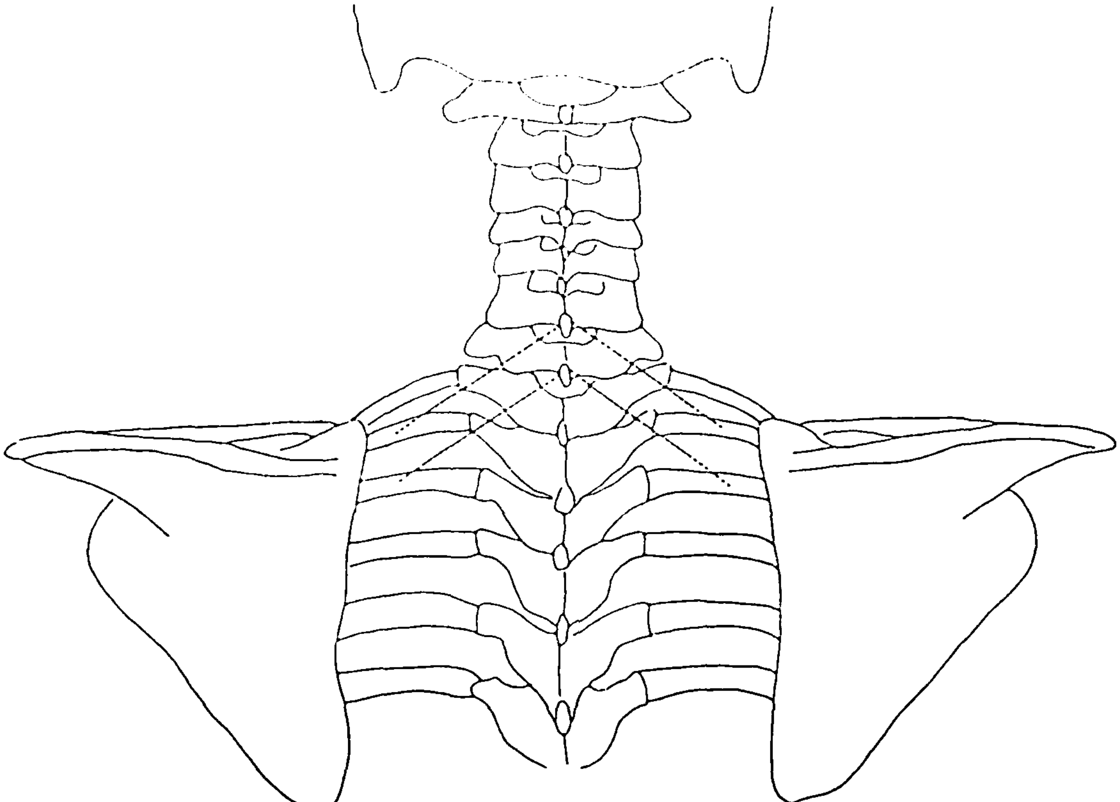

SERRATUS POSTERIOR SUPERIOR

MUSCLE NO. 20	SERRATUS POST. SUP.			
STRAND NO.	ID NO.	ORIGIN	INSERTION	AREA(CM**2)
118	2001	1604	711	0.3
119	2002	1605	811	0.3

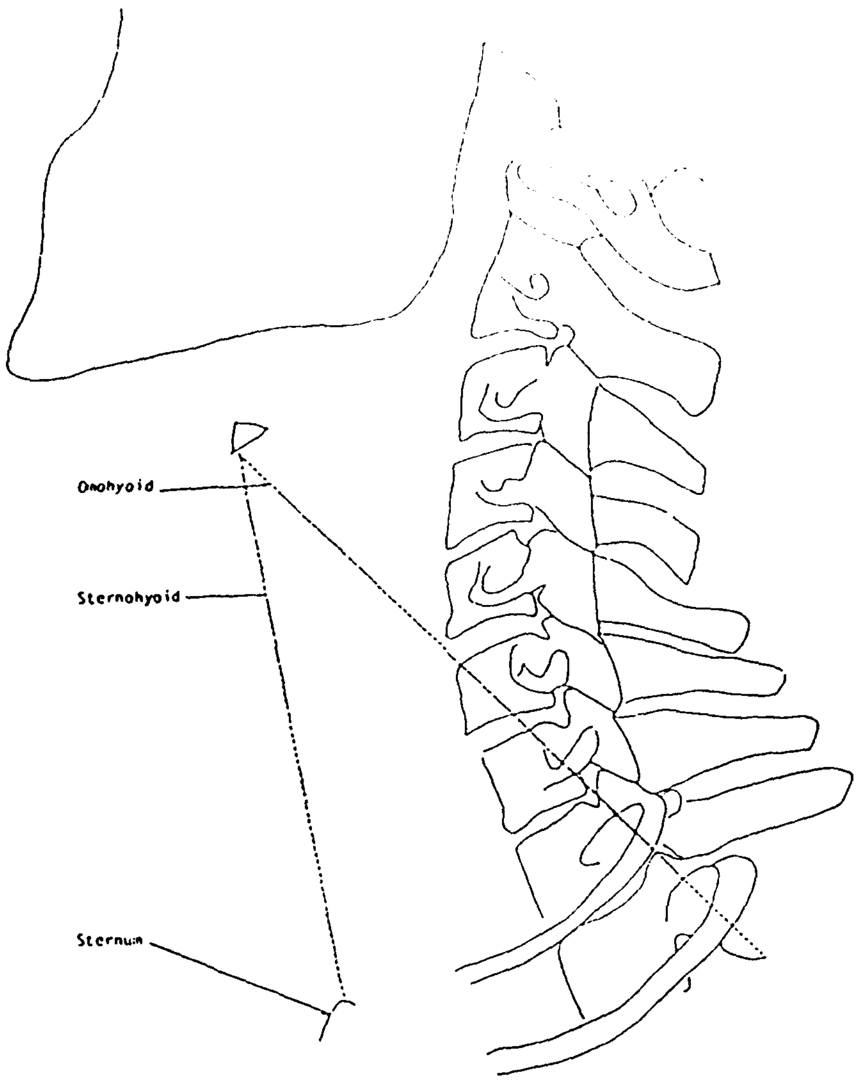
Figure A.20 Vector description of the serratus posterior superior

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 will be 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 us to simplify 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 unacceptable 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.



MUSCLE NO. 21	STERNOHYOID			
STRAND NO.	ID NO.	ORIGIN	INSERTION	AREA(CM**2)
120	2101	1501	1701	2.4
MUSCLE NO. 22	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

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