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Development of the muscle system for an Omni-Directional Dummy neck

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DIPLOMA'S THESIS

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To my family

ABSTRACT

Neck injuries in car accidents have became a serious problem over the last years. Car accidents can results in non-physiological motion in the neck and lead to injuries even at low collision speeds. These injuries are usually soft tissue injuries and are denoted whiplash injuries. Whiplash victims suffer from symptoms, which include pain in neck/head, upper thoracic spine and limited range of motion. Even if whiplash is not life threatening it can lead to long term consequences such as disability, sick leave and lost work productivity. This in turn means a great loss to the individual as well as the society. During recent years new techniques and methods for reducing whiplash injuries in the automotive field have been introduced. However, they some limitation: they just focus on rear-end impact and does not take into account the other impact modes in which whiplash injuries occur. Further investigations done have showed that other impacts such as frontal or frontal oblique may be as large a problem as rear-end impact is. Due to the relationship established between the rear-end collision and the whiplash investigation most developed dummies designs were only able to perform flexion-extension motion but it is important to consider the lateral movements involved in most car accidents.

The objective of this research was to continue and improve the development of a previous Omni-Directional Dummy neck prototype (ODD neck II), studying the human muscle system and looking for the possible solution for to replicate its. It should reproduce with a humanlike behaviour in order to make the dummy feasible to use on any kind of impacts: lateral, oblique, roll-over, far side, frontal and rear-end. Consequently, the prototype will contribute to a better understanding of injury mechanisms.

Cables moved by pneumatic, hydraulic cylinders and motor linear actuators have been considered to replicate the muscle system. This is only a theoretical work that aims to give some ideas for the development of the muscle system in dummy neck. It is possible to use this thesis as a good basis for further studies.

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CONTENTS

1. INTRODUCTION

In a car-to-car accidents the car occupants body is exposed to forces of a greater magnitude then the muscles can counteract. That can result in non-physiological motion and lead to injuries even at low speeds. These injuries are responsible for one third of all automotive injuries leading to permanent disability. One of these injuries is the whiplash injury. Whiplash Associated Disorders (WAD) is generic term for neck injuries with symptoms like pain in neck/head and upper thoracic spine, limited range of motion and a variety of poorly defined symptoms (Eck J. C.,2001). Although classified as minor injuries, they can have serious consequence for the patient's professional and personal life, due to often long-lasting symptoms. Nygren (1984) found that 18% of all accidents in Sweden involving injured drivers were rear end collisions and similar findings have been reported by others (States et al., 1972; James et al., 1991). Data published by Langwieder et al. (1981) and Kahane (1982) indicate that 80%-90% of those injuries in rear impacts sustained neck injuries. In USA the number of whiplash injured are approximately 4 per 1000 citizens per year. Between 4% and 42% of patients with accident related neck injuries report symptoms several years later (Eck J. C.,2001). The comparison of major accident samples from the German Motor insures shows that the incidence of WAD in Motor Vehicle Accidents has almost double in the last 20 years (Hell, W.;1999).Whiplash injury due to car collisions is one of the most aggravating traffic safety problems with serious implications for the European society. Yearly more than a million European citizens suffer neck injuries implying tremendous societal cost roughly at least 10 Billion Euro (Svensson, M.;2002). WAD are a major concern for the modern society in terms of both injuries and cost. A lot of work should be taken in trying to quantify the problem and determine means of injuries and cost reduction.

During recent years new techniques and methods for reducing WAD in the automotive field have been introduced. Many of these developments focus on rear-end impact only and do not take into account the other impact modes in which whiplash injuries occur. Further investigations (Cappon, .H.,2003) done have showed that frontal impact may be as large a problem as rear-end impact is, even though injury risk are lower. Inside the frontal accidents with WAD, the frontal oblique collisions show the highest incidence of injury (60-70% of the frontal accidents are offset (Friedman D.,1993)) and the impact severity in frontal accidents is significantly higher than in rear-end accidents.

In order to predict the behaviour of the human body during impact in car accidents several methods are used. Crash test with dummies, volunteers, Post Mortem Human Subjects (PMHS) and mathematical models are different ways to approach the problem. Data taken from PMHS and volunteer test are applied to validate crash test dummies and mathematical models. The aim is to make the dummy respond in a human like way. The validated dummy is a repeatable and reproducible instrument that can be used in the development of safer cars.

Due to relationship established between the rear-end collision and the whiplash investigation most dummies designs developed were only able to perform flexion-extension motion but it is important to consider the lateral movements involved in most of car accidents.

1.1 OBJECTIVE

The purpose of the first part of this thesis is to do a background of the most important properties and characteristics of the human muscle system, and analysis of the mathematical modelling to understand the behaviour of the neck in different types of impacts and speeds.

The purpose of the second part is to make components that replicate the muscle system, and in the end to think about the different possible solution to move the cables with human neck motion.
This thesis is a theoretical studying to give some solutions and ideas about how to reproduce the human muscle system. There isn't time to do dynamic analysis with dynamic programs like Adams and Madymo or real tests, to understand which will be the best result.

1.2 BIOMECHANICS OF THE HUMAN CERVICAL SPINE

In the process of developing a dummy neck, it is necessary to identify the essential parts of the neck and therefore, knowledge about the human anatomy is needed. The basic mechanical elements of the neck are the vertebrae, the intervertebral discs, the muscles, the ligaments, the vein system and the spinal cord. In this chapter the fundamentals of the human cervical spine will be presented.

1.2.1 Functional anatomy of the spine

Vertebrae. The cervical spine has seven vertebrae, referred as C1 to C7 from the top, which constitute the neck. (Fig. 1.1) The main roles of this upper spine segment are to hold the head up, allow its different movements and protect the spinal cord. Among these seven vertebrae, the first and second vertebrae, atlas and axis, are distinct from each other and from the lower five vertebrae, which are similar in shape and functionality.

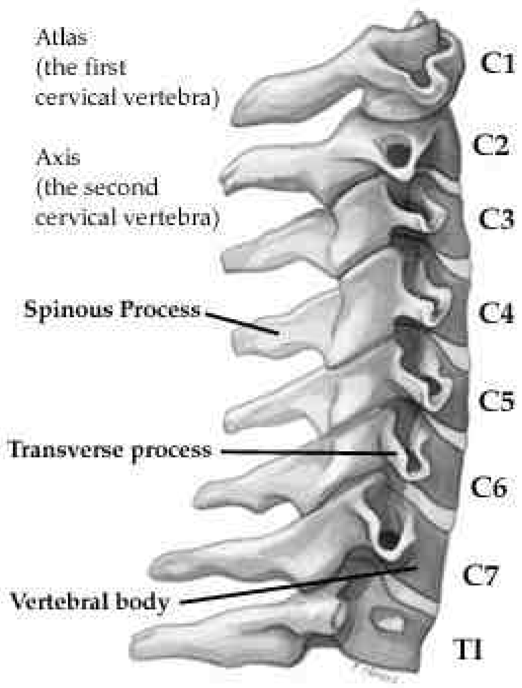


Fig. 1.1 Neck structure (adapted from Hughston Sports Medicine Foundation)

Due to these differences, the cervical vertebral column can be divided into the lower and the upper cervical spine.

The upper part of the cervical spine consists of the two upper most vertebrae, C1 and C2. These two vertebrae differ in their structure compared to the five other vertebrae (C3-C7) in the lower cervical vertebrae column.

The first vertebra C1 is called “the atlas” (Figure 1.2) and can be considered as a ring of bone; it has no body and no spinous process. The atlas is provided at the superior side with a pair of facets, each covered with cartilage, which articula te with the bases of the skull, the atlanto-occipital joint. The skull part of this articulation is formed by the condyles of the occipital bone referred to as occipital condyles. This articulation allows a nodding motion of the head (flexion and extension motions). Its vertebral foramen is bigger than those of other vertebrae is.

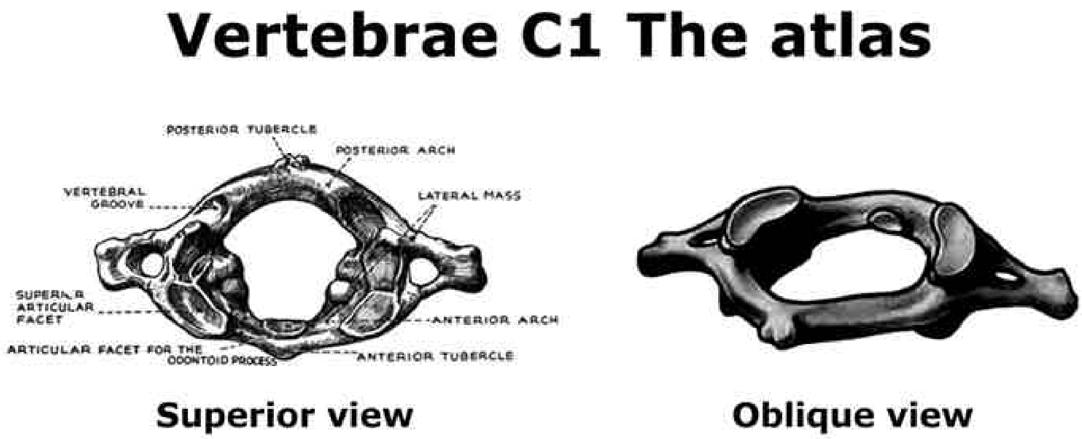


Fig. 1.2 Atlas (adapted from Human Anatomy&Physiology[5])

C2 is called “the axis” (Figure 1.3) and has a knoblike dens, the odontoid process, projecting superior from its body. This dens fits into the vertebral foramen of the atlas. This atlanto-odontoid joint allows horizontal rotation in the upper part of the cervical spine, C1 rotates on top of C2 with the dens as the centre of rotation.

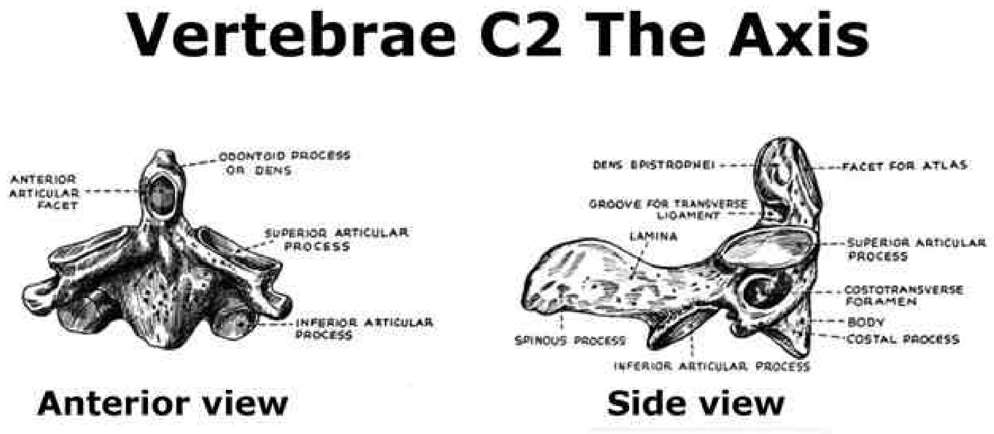


Fig. 1.3 Axis (adapted from Human Anatomy&Physiology [5])

The lower cervical region consists of vertebrae C3 through C7 (Fig. 1.4 and 1.5), each of which comprises of a cylindrically shaped body and an arch. These enclose the vertebral foramen which forms the spinal canal through which the spinal cord passes. The arch includes two pairs of articular facets, a spinous process and two transverse processes. The processes constitute attachment points for muscles and ligaments. The posterior spinous process will come in contact when the neck performs hyperextension, extreme movement in extension, and the transverse processes will behave closely on lateral bending.

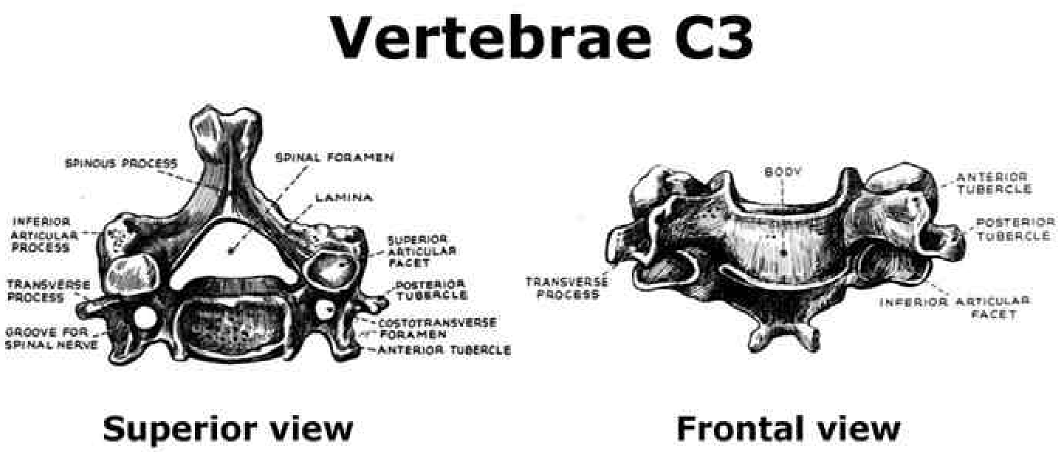


Fig. 1.4 C3 (adapted from Human Anatomy&Physiology [5])

The articular facets are almost flat, covered with cartilage and obliquely oriented with different values of angles (Fig. 1.5) from one vertebra to another but with a mean backward angle of about 42 degrees in the sagittal plane.

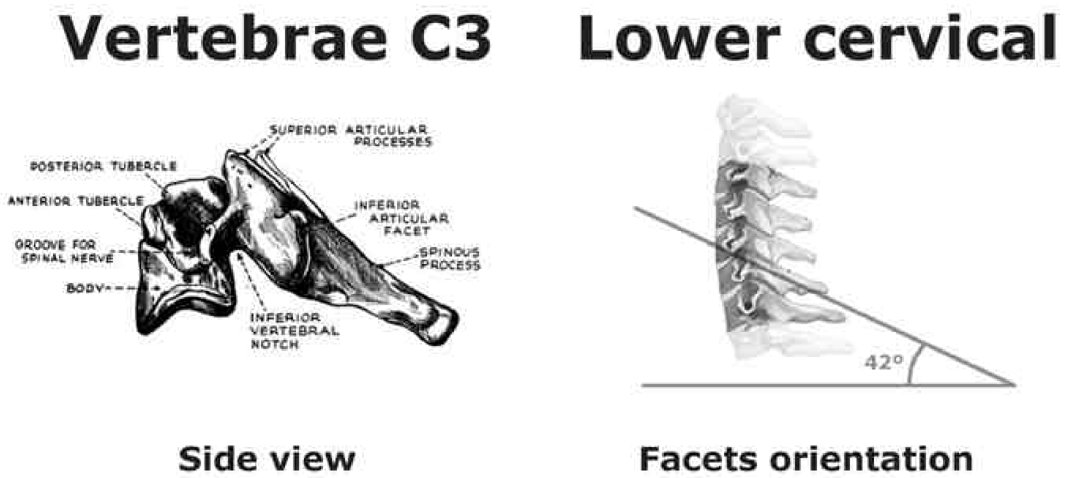


Fig. 1.5 C3 side view and lower spine facets orientation (adapted from Human Anatomy&Physiology [5])

Intervertebral discs. They are placed between two adjacent vertebrae except concerning the occipito-atlanto-axial region. The discs are a fibrocartilaginous joints that allow movement in all directions and work as a damper. They are thicker anterior than posterior and provide the cervical spine an anteriorly convex curve, called cervical lordosis. Due to their composition, the discs are more sensitive to rotational strain than to compression, tension and shear (Fig. 1.6).

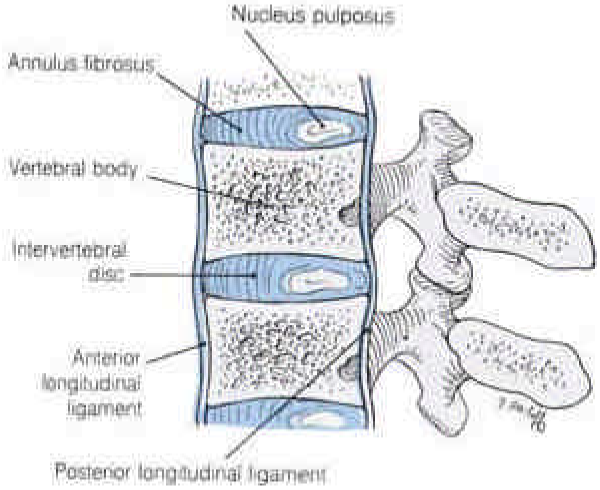


Fig. 1.6 Intervertebral disc

The spinal cord. The spinal cord runs inside the spinal canal (Figure 1.7). It consists of the grey and white matter, which can be considered as semi-fluid cohesive masses. The grey matter is shaped like a butterfly and is surrounded by the white matter. It consists of a mixture of neurone cell bodies. The white matter consists of nerve fibres, which allow communication between different parts of the spinal cord or between the spinal cord and the brain. It is covered by the pia matter, which is again surrounded by the arachnoid. The spinal cord cannot move up and down axially within the canal but it adapts itself to the changing of lengths during motions of the spine by plastic deformation.

Dorsal nerve roots are leading from the spinal cord to the spinal ganglion, which is situated in the intervertebral foramen.

Both spinal cord (covered by pia matter) and the spinal ganglions are surrounded by the Cerebro Spinal Fluid (CSF) and covered by dura matter.

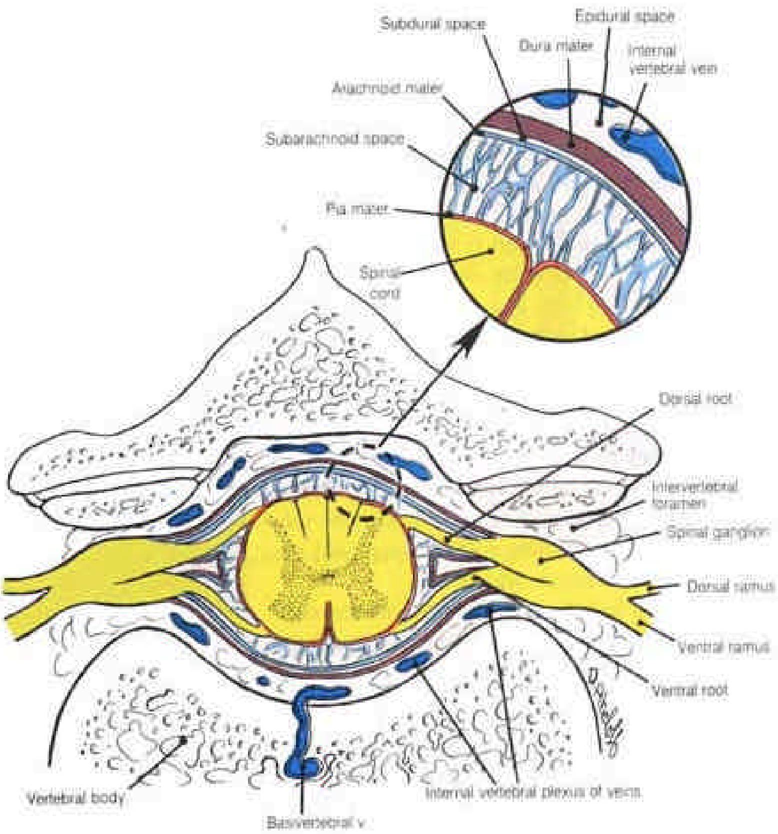


Fig. 1.7 A horizontal cross-section of a cervical vertebra with soft tissues.

Ligaments. . Ligaments allow spinal motion within physiologic limits and prevent excessive motion to protect the spinal cord. They connect adjacent vertebrae or extend over several vertebrae. Ligaments resist tension and fold in compression.
Figure 1.8 on the left depicts the ligaments of the lower cervical spine. The anterior and posterior longitudinal ligaments are strong ligaments attached to the anterior and posterior parts of the vertebral bodies and intervertebral discs. The interspinous ligament is a thin, tough membrane between adjacent spinous processes and the flaval ligament is strong band between adjacent laminae. The capsular ligaments enclose the facet joint. The nuchal ligament (not shown) is a thin and weak triangular membrane joining all cervical spinous processes and interspinous ligaments to the posterior midline of the occiput; its posterior edge runs from the occiput to the spinous processes of C_7.
The anterior longitudinal ligaments limits extension of the spine. The other ligaments are all posterior ligaments and limit flexion.
The most important ligaments of the upper cervical spine are shown in Figure 1.8 on the right the tectorial membrane is a broad and strong band which extends the posterior longitudinal ligament from the axis to the occiput. The anterior and posterior occipito-atlantal and atlanto-axial membranes are continuations of the anterior longitudinal and flaval ligament, respectively. The transverse ligament is a strong horizontal ligament of the atlas,

which holds the dens against the anterior arch of the atlas to constrain the dens posteriorly. The dens is held anteriorly by the apical and alar ligaments. The apical ligament run from the top of the dens to the occiput. The alar ligaments extend from the dens on each side of the apical ligament to the medial side of the occipital condyles and the atlas. Apical and alar ligaments limit rotation of the upper cervical spine

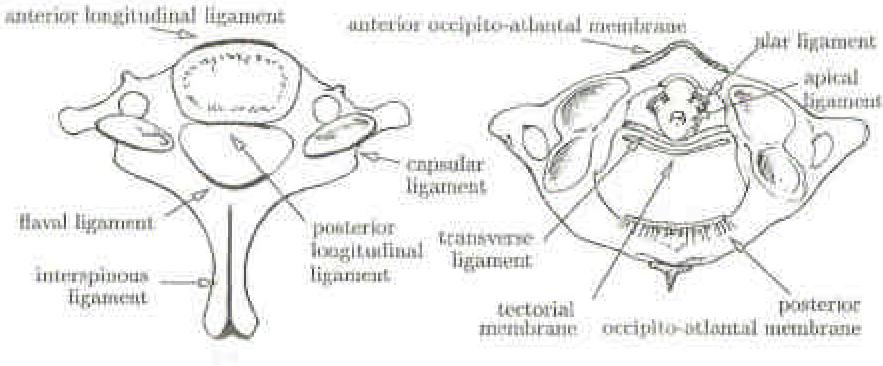


Fig. 1.8 Lower (on the left) and upper (on the right) cervical ligaments (Adapted from Sances)

The vein system. The vertebral vein system is a low-pressure system. It is divided in three different parts, which intercommunicate with each other: the internal vein network, the in- and out- leading bridging veins and the external veins plexus. The internal vein network is the largest one, surrounding the dura matter. It consists of two-vein network (plexus), which are situated posterior and anterior within the spinal canal. The two plexus (posterior and anterior) are joined through several vessels respective small networks. The volume capacity of these plexus is about 100 ml or even more. This is
20 times the arterial capacity; it is much larger than required to return the blood brought in by the arteries. It seems to be clear that bringing out blood is not the main job of these plexus. More probable they serve as regulator to balance the volume and pressure changes during movements of the cervical spine and to the storage of blood. Out of this reason, they do not have any valves. The blood is able to flow in any direction within the plexus.

A similar vein plexus is outside the cervical spine, the external vertebral venous plexus (anterior and posterior) The intervertebral veins (Venae Intervertebralis), which lead in groups through the intervertebral foramen on both sides of the vertebral body, and the basivertebral veins (Venae Basivertebralis), which lead in radial direction through the vertebral body (Figure 1.9 and Table 1.1).

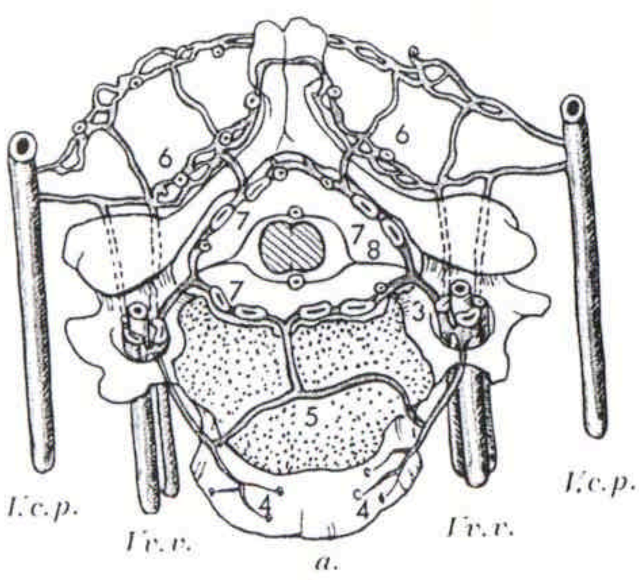


Fig. 1.9 Charting of the internal and external cervical venous plexus: a= Cevical vertebrae; Vv.v= Venae vertebralis; V.e.p.= Venae cervicalis profunda; 3= Venae intervertebralis; 4= Anterior external cervical venous plexus; 5= Venae basiventralis; 6= Posterior external cervical venous plexus; 7= Internal cervicalvenous plexus

Vein	Diameter in mm
Internal vertebral cervical venous plexus, anterior	2.0-2.5
Internal vertebral cervical venous plexus, posterior	2.5-3.0
Intervertebral cervical veins	<5.0
Basivertebral cervical veins	1.0-2.0
External vertebral cervical venous plexus, anterior	0.1-0.5
External vertebral cervical venous plexus, posterior	0.2-0.5

Table 1.1 Diameters of cervical vertebral veins (Clemens, 1961)

1.2.2 Human muscle system of the neck

Muscles is an important component for the stability and the flexibility of the neck, because the spine with its ligaments intact but devoid of the muscle, is an extremely unstable structure.

The muscles and complex neuromusculature controls are required to provide stability of the trunk in a given posture and to produce movements during physiologic activity. The muscles may also play a role in protecting the spine during trauma in which there is a time for voluntary control and possibly in the post-injury phase.

The muscles exist in several layers and they come in different lengths

The head is in equilibrium (Fig. 1.10) when the eyes look horizontally.
The head taken as a whole constitutes a lever system:
- the fulcrum O lies at the level of the occipital condyles;
- the force G is produced by weight of the head applied though its centre of gravity lying near the sella turcica;
- the force F is produced by posterior neck muscles which constantly counterbalance the weight of the head, which tends to tilt it forward.

This anterior location of the centre of gravity of the head explains the strength of the posterior neck muscles relative to the flexor muscles of the neck. In fact the extensor muscles counteract gravity whereas the flexor are helped by gravity. This also explains the constant tone in these posterior neck muscles preventing the head from tilting forwards. When one sleep while sitting the tone of these muscles is reduced and the head falls on the chest.

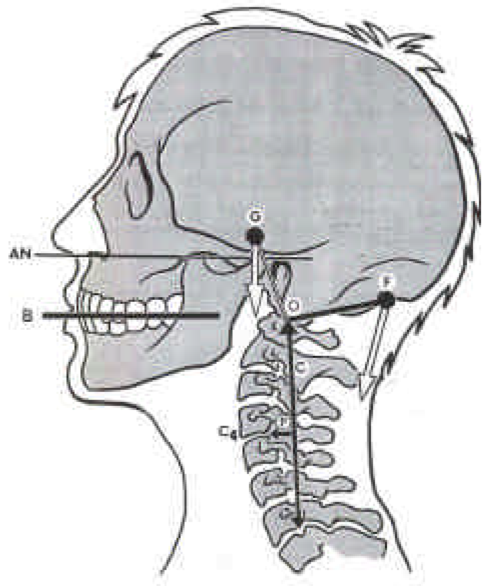


Fig. 1.10

This muscles should be called the sterno-cleido-occipito-mastoid (S.C.O.M.) as it is made up of four distinct band (Fig. 1.11.A):
- a deep band, the cleido-mastoid (C.M.) running from the medial third of the clavicle to the mastoid process;
- a superficial band, the cleido-occipital which overlies the bulk of the cleido-mastoid and is inserted into the lateral third of the superior nuchal line of the occipital;
- a superficial band, the sterno-occipital and the sterno-mastoid, both of which arise by a common tendon from the superior margin of the sternum. The sterno-occipital is inserted along with the cleido-occipital into the superior nuchal line, while the sterno-mastoid is insert into the superior and anterior borders of the mastoid.

Unilateral contraction of this muscle (Fig. 1.11.B) gives rise to a triple movement combining rotation of the head contralaterally, lateral flexion ispilaterally and extension.
The effects of bilateral contraction vary according to the state of contraction of the other neck muscles:
- if the cervical vertebral column is flexible, this bilateral contraction accentuates the cervical lordosis with extension of the head and flexion of the cervical column (see Fig. 1.20.C);

- if on the contrary, the cervical column is kept straight and rigid by the contraction of the prevertebral muscles, then bilateral contraction produces flexion of the cervical column relative to the thoracic column and forward flexion of the head (see Fig. 1.21.C).

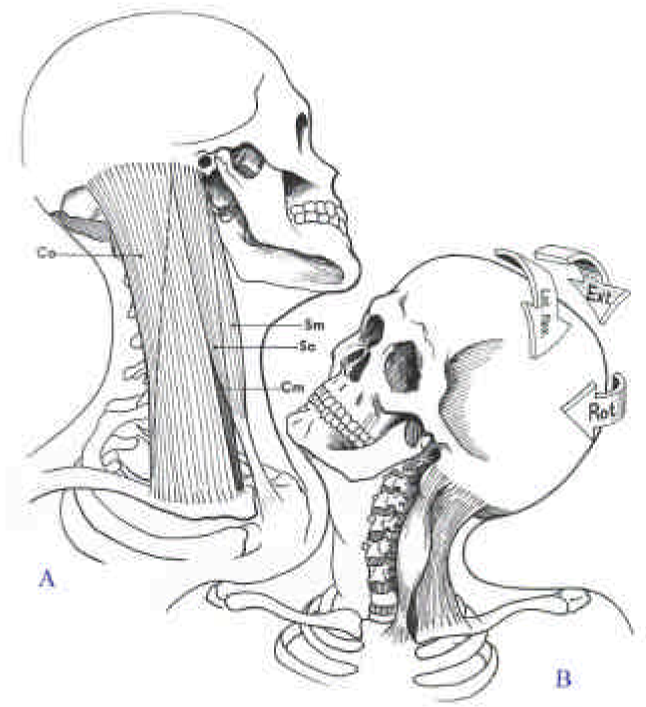


Fig. 1.11

The prevertebral muscles:

1) the longus cervicis (l.c.) is the deepest of the prevertebral muscles (Fig.1.12) and runs on the anterior surface of the column from the superior arch of the atlas to the third thoracic vertebra. It consists of three sets of fibres atomically:

- an oblique descending set attached to the anterior tubercle of the atlas and to the anterior tubercle of the traverse process of C_3-C_6 by three or four tendinous slips;
- an oblique ascending set attached to the bodies of T_2 and T_3 and to the anterior tubercle of the transverse process of C_4-C_7 by three or tendinous slips;
- a longitudinal set lying deep to the former two sets and just lateral to the midline. It is attached to the bodies of the first three thoracic vertebrae and the last six cervical vertebrae.

Thus the longus cervicis, on the either side of the midline, covers the whole anterior surface of the cervical column. When both muscles contract symmetrically they flatten the cervical curvature and flex the neck.

Contraction of one longus cervicis produces forward and lateral flexion of the cervical column ipsilaterally.

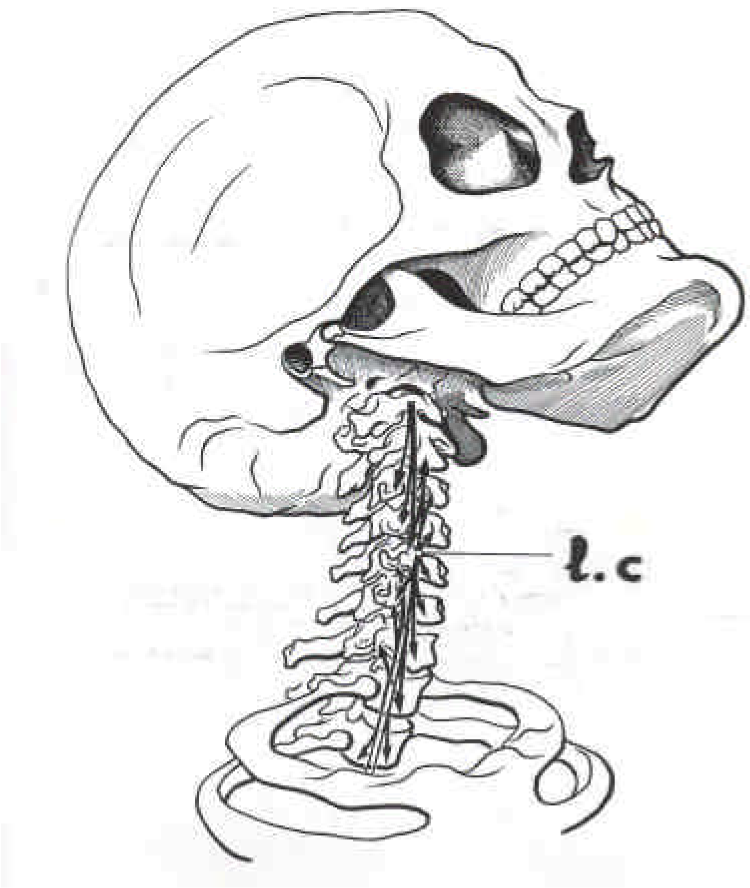


Fig. 1.12

2) the anterior and lateral rectus muscles belong to the upper reaches of the cervical vertebral column (Fig.1.13).
The rectus capitis anterior consists of two bans. The deep band (also called the rectus capitis anterior maior R.A.Ma.) is attached to the inferior surface of the basi-occiput anterior to the foramen magnum and to the anterior tubercles of the transverse processes of C_3-C_6, and overlies the longus. It acts on the suboccipital vertebral column and on the upper part of the lower cervical column. Bilateral contraction produces flexion of the head on the cervical column and a flattening of the lordosis of the upper portion of the cervical column. Unilateral contraction causes forward and lateral flexion of the head ipsilaterally.
The superficial band (also called the rectus capitis anterior minor R.A.Mi) lies posterior and lateral to the former and stretches from the basis-occiput to the anterior aspect of the lateral mass of the atlas up to the anterior tubercle of its transverse process. It runs obliquely inferiorly and laterally. Bilateral contraction of these muscles produces flexion of the head on the cervical column at the level of the atlanto-occipital joint. Unilateral contraction produces a triple movements combining forward flexion, rotation and lateral flexion ipsilaterally.
The rectus capitis lateralis (r.l.) is the highest of the intertransverse muscles and is attached to the jugular process of the occiput to the anterior tubercle of the transverse process of the atlas. It lies lateral to the anterior rectus and overlies the anterior surface of the atlanto-occipital joint. Its bilateral contraction produces flexion of the on the cervical column, its unilateral contraction produces a slight lateral flexion of the head ispilaterally.

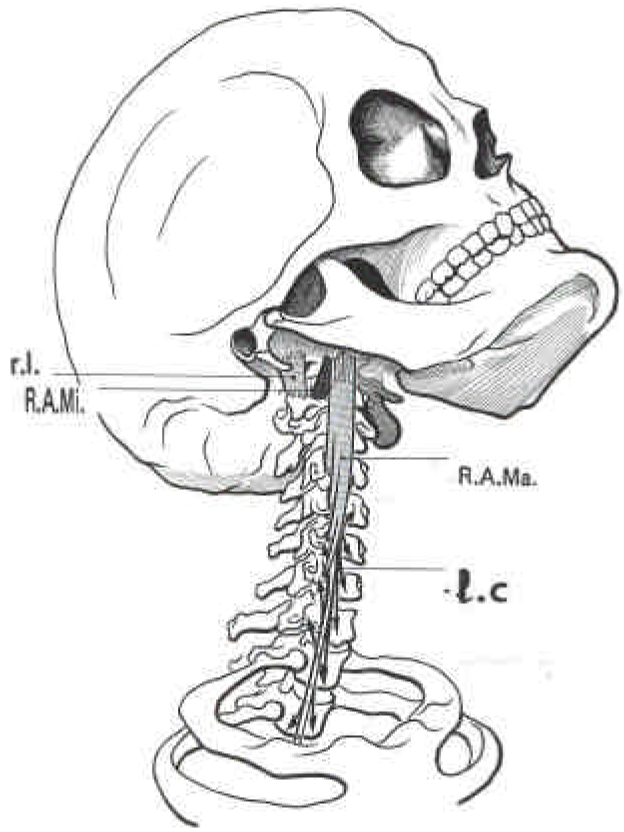


Fig. 1.13

3) the three scalene muscles lie on the antero-lateral aspect of the cervical column like muscular tighteners (Fig.1.14) and connect the transverse processes of the vertebrae to the first two ribs.

The scalenus anterior (s.a.) is triangular with its apex lying inferiorly and takes origin from the anterior tubercles of the transverse processes of C_3-C_6. Its fibres converge down to be inserted by tendon into the scalene tubercle on the superior aspect of the first rib.

The scalenus medius (s.m.) lies in contact with the deep surface of the former and arises by tendinos slips from the posterior tubercles of the transverse processes of C_2-C_7, the lateral edge of the gutter-like depression in the transverse processes of C_2-C_6 and the transverse process of C_7. Its fibres run obliquely, inferiorly, laterally and slightly anteriorly to be inserted into the superior surface of the first rib.

The scalenus posterior (s.p.) lies posterior to the other two. It arises by thee tendinous slips from posterior tubercles of the transverse processes of C_4-C_6. It is inserted by a flat tendon into the superior border and the lateral aspect of the second rib.

When these muscles contract symmetrically the cervical column is flexed on the thoracic column and the cervical lordosis is accentuated if the neck is not held rigid by contraction of the longus cervicis. If the scalenes contract only on one side the cervical column is laterally flexed and rotated toward the side of their contraction.

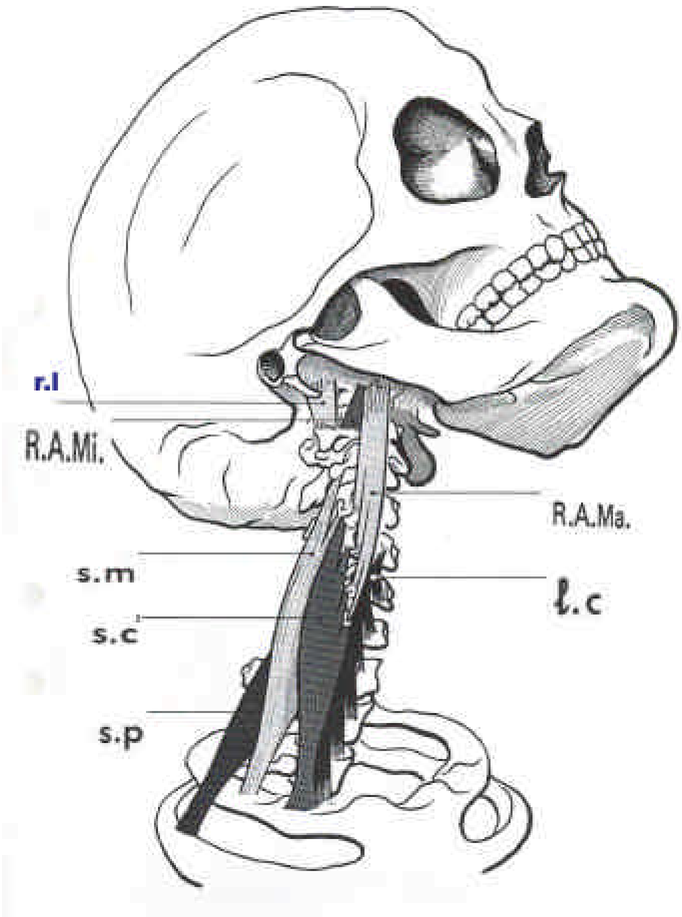


Fig. 1.14

The prevertebral muscles viewed as a whole (Fig.1.15)
Flexion of the head on the cervical column and flexion of the neck on the thoracic column depend on the anterior muscles of the neck.

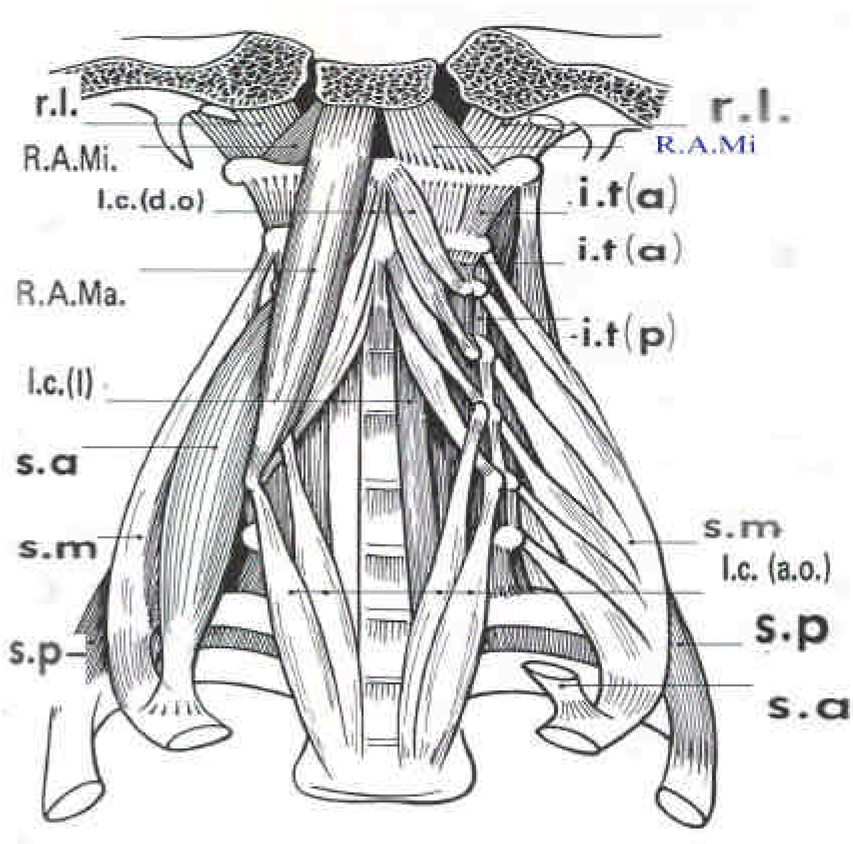


Fig. 1.15

The posterior muscles of the neck are made up of four muscular plane superimposed on one another, as follows (Fig. 1.16):
- the deep plane, directly adherent to the vertebral bones and joints, containing the intrinsic muscles of the suboccipital vertebral column running between the occiput, the atals and axis, rectus capitis posterior major (1) and minor (2), the obliquus capitis superior (3) and inferior (4), the cervical transverso-spinalis (5), the interpinous muscles (6);
- the semispinalis plane contains the semispinalis capitis (7) and semispinalis cervicis (8), laterally, the transversus thoracis, the longissimus thoracis and the superior portion of the iliocostalis (11);
- the plane of the splenius and the levator scapulae ban be subdivided into two layers, the splenius capitis (9), the spenius cervicis (10), the levator scapulae (12);
- the superficial plane is comprised of the trapezius (15), the sterno-cleido mastoid which belongs to the posterior neck only in its postero-superior part (14 and 14').

In the depth of the figure, though the gap between the muscles one can see the attachments of the scalenus medius and posterior (13).

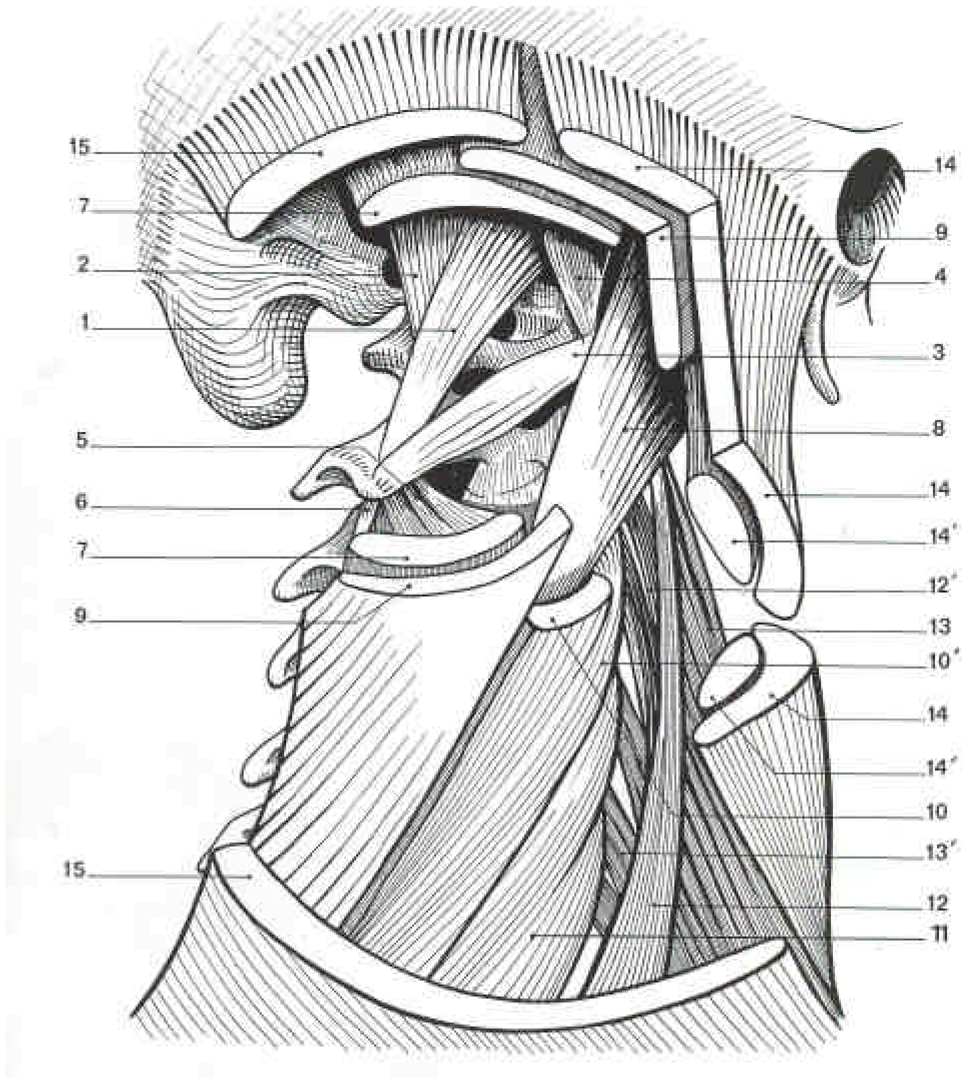


Fig. 1.16

The suboccipital muscels can be seen from the back (Fig.1.17.B) and from the side (Fig.1.17.A). The figures show:

- the rectus capitis posterior major (1), which is triangular in shape with its based located superiorly. It is attached to the spinous process of the axis and the inferior nuchal line of the occiput;
- the rectus capitis posterior minor (2), which is triangular, flattened and smaller than the major to which it lies deep. It lies closer to the midline and is attached to the tubercle of the posterior arch of the atlas and to the medial third of the inferior nuchal line of the occiput;
- the obliquus capitis inferior (3), which is an elongated, thick and fusiform muscle lying inferior and lateral to the rectus capitis major. It is attached to lower border of the spine of the axis and to the posterior margin of the trasverse process of the atlas;
- the obliquus capitis superior (4), which is a short, flattened triangular muscle lying posterior to the atlanto-occipital joint. It is attached to the transverse process of the atlas and to the lateral third of the inferior nuchal line of the occiput.;
- the interspinous muscles (5) lie on either side of the midline between the spines of C_3-C_7.

The obliquus capitis inferior, as a result of its location, is important in maintaining the integrity of the atlanto-axial joint at rest and during movements. This muscle pulls back the

transverse processes of the atlas and so, when the two obliques contract symmetrically, they pull back and extend the atlas on the axis.
Unilateral contraction of the four suboccipital muscles produces lateral flexion of the head ispilaterally at the atlanto-occipital joint.
Bilateral simultaneous contraction of the suboccipital muscles produces extension of the head on the upper cervical column at the atlanto-occipital joint, the muscles involved being the rectus posterior minor and the superior oblique. At the atlanto-axial joint the muscles involved are the rectus posterior major and the inferior oblique.
In addiction to extension and lateral flexion, these muscles also produces rotation of the head. Let us first consider the atlanto-occipital joint, it is clear that superior oblique produces a 10° rotation of the head contralaterally. Let us now consider the atlanto-axial joint, it is clear that contraction of the rectus posterior major and the inferior oblique produces rotation of the head ipsilaterlly.

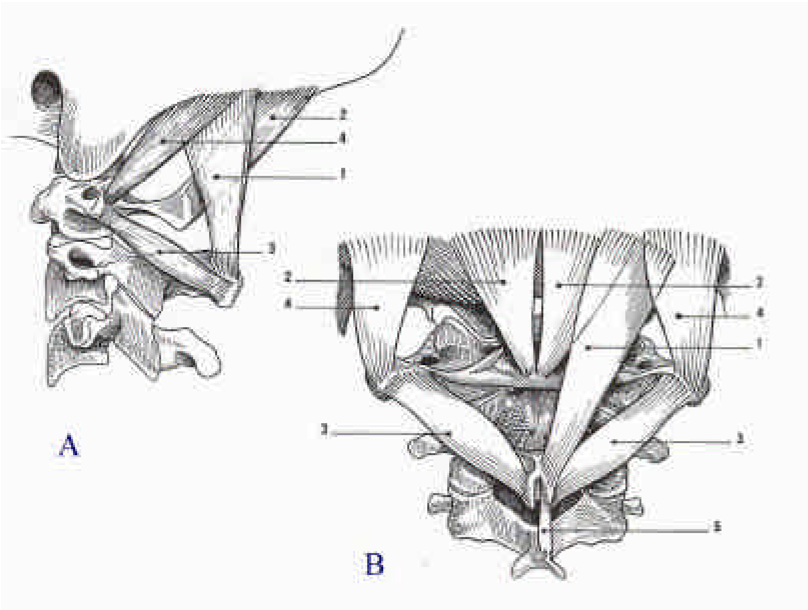


Fig. 1.17

The deep plane consists of the suboccipital muscles in the upper cervical region and of the transversospinalis muscles in the lower cervical region. These muscles are closely adherent to the vertebrae and fill the groove formed by spines, the laminae and the transverse processes. Their contraction produces when the muscles contract bilaterally and symmetrically, extension of the cervical column and accentuation of the cervical lordosis (therefore they are equivalent to an erector pinae cervicis), when only one muscle contracts, lateral flexion ispilaterally and rotation of the cervial column contralaterally (therefore it is analogous to that of the sternomastoid on the head).

The superficial plane of the posterior neck muscles (Fig.1.18.A) consists of the trapezius (Tr), which arises fanwise from a line joining the medial third of the superior nuchal line of the occiput, the posterior cervical ligament and the spines of the cervical and thoracic vertebrae town T_{10}. When both trapezius muscles contract symmetrically, the cervical column is extended and the cervical lordosis is accentuated. When this extension is counterbalanced by the antagonistic action of the anterior muscles of the neck, they act as tighteners that stabilise the whole cervical column. When only the trapezius contract (fig.1.18.B) the head is extended and the cervical lordosis is accentuated while it is flexed ipsilaterally and rotated contralaterally.

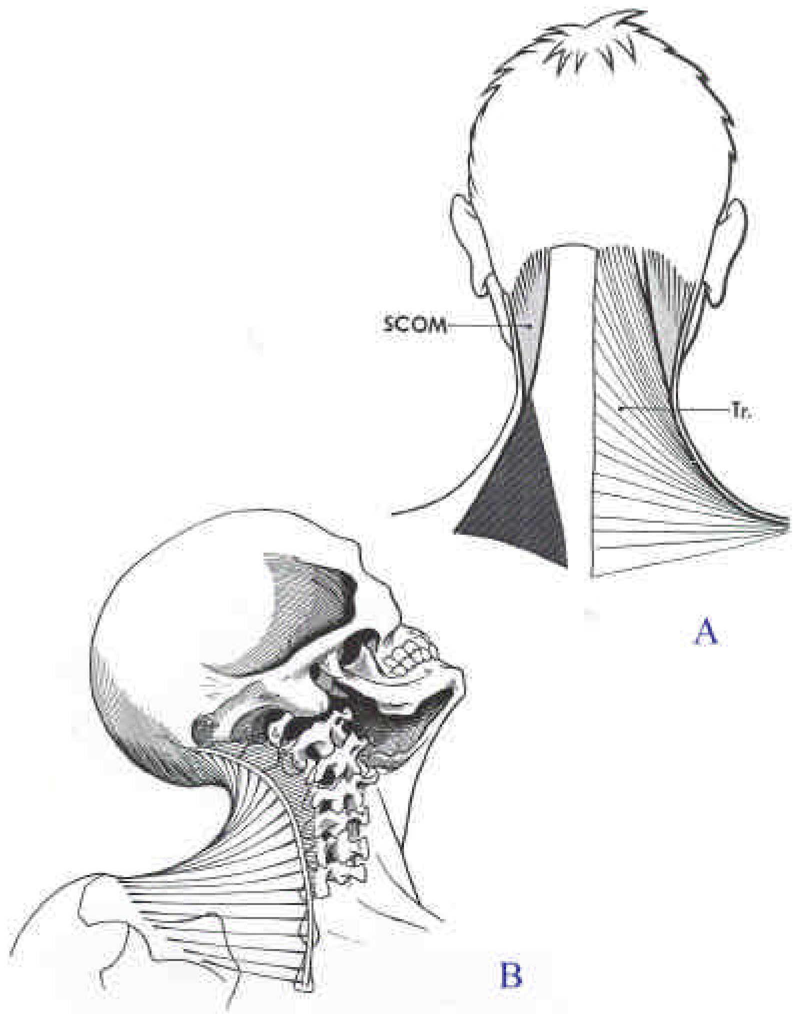


Fig. 1.18

Deep to the trapezius is the third muscular plane (Fig. 1.19.A) comprised of the splenius and the levator scapulae.
The splenius, which runs from the skull to the thoracic region, arises from the lower six cervical spines, the posterior cervical ligament, the upper four thoracic spine and the interspinous ligament. These muscles can to be inserted as two distinct bands:
- the cephalic band, the splenius capitis (9), is inserted into the lateral half of the superior nuchal line of the occiput inferior to the sterno-mastoid and into the mastoid process;
- the cervical band, the splenius cervicis (10), is inserted into the transverse processes of the upper three cervical vertebrae.
When these muscles contract bilaterally and symmetrically they extend the head and accentuate the cervical lordosis. When only one muscle contracts, asymmetrically, it produces the triple movement of extension, lateral flexion and rotation ipsilaterally.
The levator scapulae (12), lying lateral to the splenius cervicis, shares its origin from the transverse processes of the upper four cervical vertebrae. When the levators contract bilaterally and symmetrically they extend the cervical column and accentuate the cervical lordosis. When only one levator contract, asymmetrically, it produces like the splenius the triple movement of extension, lateral flexion and rotation ipsilaterally.

The second muscular plane is made up of (Fig.1.19.B):

- the semispinalis capitis (7) arises inferiorly from inferiorly from the transverse processes of the upper six thoracic vertebrae, from the base of the transverse processes of the lower four cervical vertebrae and from the spines of C_7 and T_1. It is inserted into the squama of the occiput and the medial half of the area between the two nuchal lines. When the semispinalis muscles contract bilaterally and symmetrically they extend the head and the cervical column and accentuate the cervical lordosis. When one semipinalis contracts asymmetrically, it produces extension with minimal lateral flexion ipsilaterlly;
- the longissimus capitis (8) arises from the four lower cervical and the first thoracic transverse processes and is inserted into the posterior border of the mastoid process. When the longissimus muscles contract bilaterally and symmetrically they extend the head, when this extension is counterbalanced by the anterior muscles of the neck, they stabilise the head laterally, acting like inverted tighteners. When one longisimus muscles contracts asymmetrically, it produces extension and lateral flexion ipsilaterlly;
- the semispinalis cervicis (11) arises inferiorly from the transverse processes of T_1-T_5 and is inserted into the transverse processes of C_3-C_7. The deepest fibres, those running between C_7 and T_1, are the shortest, and the most superficial fibres, those running between C_3 and T_5, are the longest. When these muscles contract bilaterally and symmetrically they extend the lower cervical column. When one muscles contracts asymmetrically, it produces extension and lateral flexion ipsilaterlly;
- the longissimus thoracis also belongs to the posterior neck muscles by virtue of its uppermost fibers attached to the transverse processes of the lowest cervical vertebrae. It is more or less continuous with cervical portion of the ilio-costalis (11’), which arises from the superior margin of the first six ribs and is inserted, along with the longissimus thoracic, into the posterior tubercle of C_3-C_7.

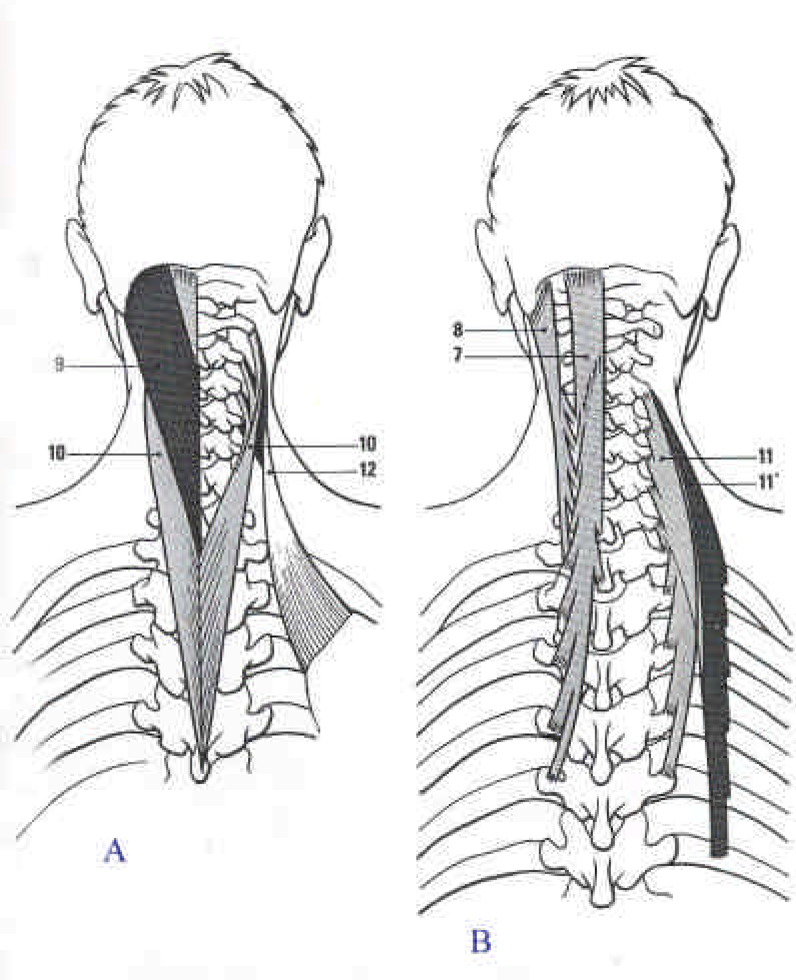


Fig.1.19

Discussion

The posterior neck muscles extend the cervical column and the head but, depending on their orientation, they can be divided into three groups:
- group I (Fig.1.20.A), comprising all muscles arising from the cervical transverse processes and running obliquely inferiorly and posteriorly to the thoracic region, the splenius cervicis (10), the semispinalis cervicis and cervical portion of the ilio-costalis (11), the levator scapulae (12). These muscles extend the cervical column and increase the cervical lordosis. When they contract unilaterally they produce extension, lateral flexion and rotation of the cervical column ipsilaterally;
- group II (Fig.1.20.B), comprising all muscles that run obliquely inferiorly and anteriorly: the transverso-spinalis (5), whih is an intrinsic muscle of the lower cervical column. The muscles linking the occiput to the lower cervical column: the semispinalis capitis (7), the longissimus capitis (8) and the splenius capitis;
- group III comprising all the muscles that brige over the cervical column without being attached to its vertebrae. They connect the occiput and the mastoid process to the scapula. They are: the trapezius (15 Fig.1.20.B), the sterno-mastoid (Fig.1.20.C), which runs diagonally across the cervical column. When the sterno-mastoids contract bilaterally and symmetrically they produce extension of the head on the cervical column (1), flxion of the cervical on the thoracic column (2) and extension of the cervical column on itself to accentuate the cervical lordosis (3).

The stability of the cervical column in the sagittal plane (Fig.1.20.D) thus depends on a constant equilibrium between: on the one hand extension by the posterior neck muscles, on the order yhe anterior and antero-lateral muscles.
The simultaneous contraction of all these muscle groups maintains the cervical column rigid in the neutral position. Thus they act like tighteners located in to the sagittal plane and in the multiple oblique planes. They are therefore essential in balancing the head and in supporting weights carried on the head.

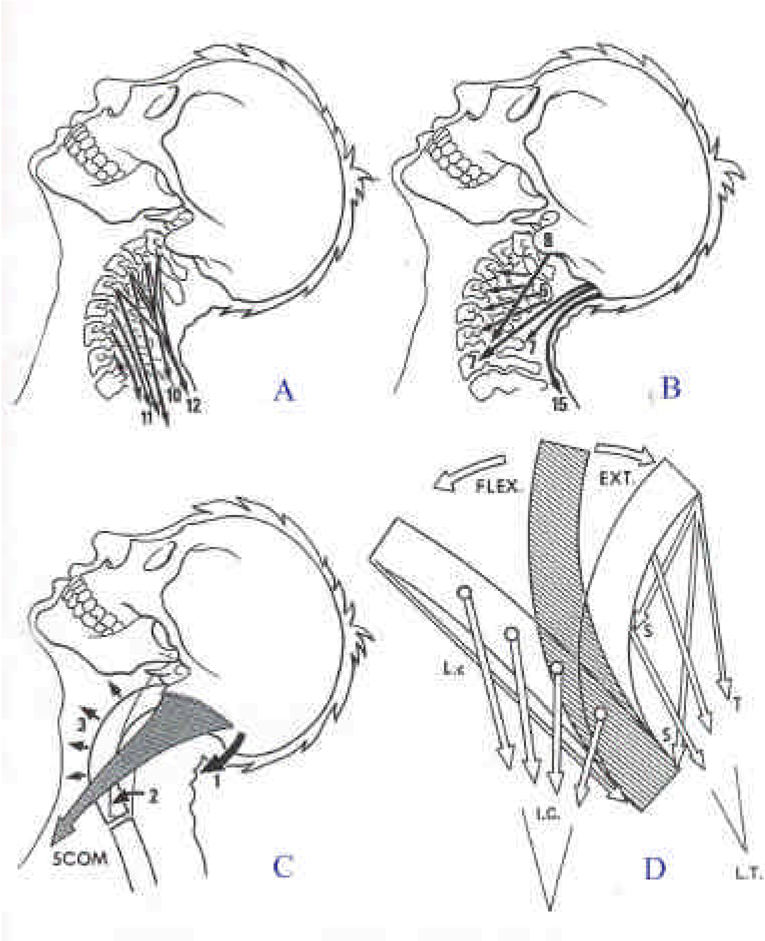


Fig. 1.20

The sterno-mastoids (SCOM) cannot by themselves steady the head and the cervical column. They need the assistance of synergistic and antagonistic muscles that set the stage by first flattening the cervical lordosis (Fig.1.21.B). They are:
- the longus capitis (L.C.) lying just anterior to the vertebrae bodies;
- the suboccipital muscles that flex the head on the cervical column (Fig.1.21.D), rectu capitis anterior and the rectus capitis lateralis;
- the supra- and- infra-hyoid muscles, which lie anterior to the cervical and act at a distance by means of the long arm of a level.

When the cervical column is held rigid, the cervical lordosis flattened (Fig.1.21.A) and extension of the head on the cervical column prevented by the anterior suboccipital muscles and the supra- and- infra-hyoid muscles, the two stero-mastoids (Fig.1.21.C) produce flexion if the cervical column on the thoracic column. Thus there is synergism-antagonism between the sterno-mastoids and the prevertebral muscles lying against the column or at some distance anteriorly.

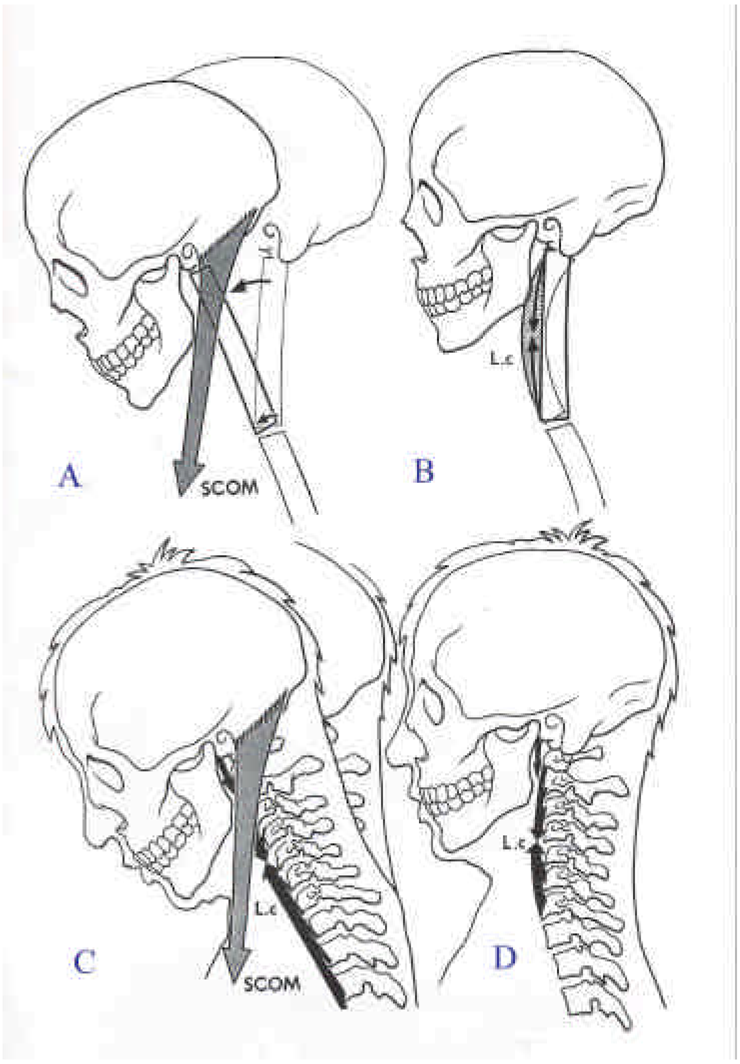


Fig. 1.21

1.3 BIOMECHANICS OF THE NECK

Biomechanics of the neck describes the motions that the neck can perform, how they are done and the range of these motions. The motions in the human neck can be defined as four movements (Fig.1.22) and the combinations of these ones. Bending forward placing the face toward the chest is flexion, bending backwards, extension. Laying the head right or left by putting the ear toward the shoulder is lateral bending and turning the head from side to side is rotation.

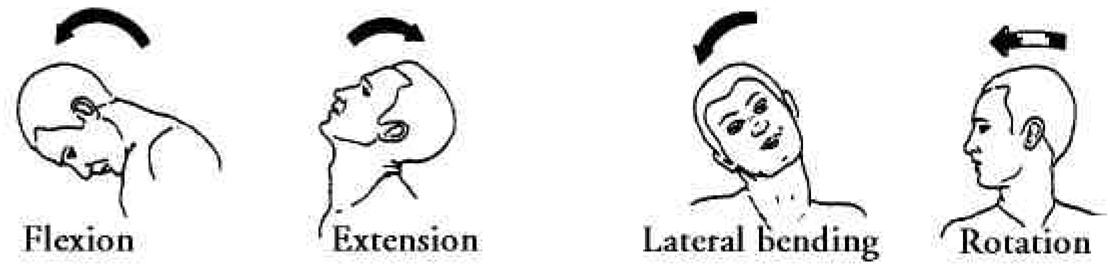


Fig. 1.22 Terms for neck motions (Adapted from Hulkey and Nusholtz, 1986)

When extension or flexion occurs, each vertebra in the cervical spine glide on the vertebra below, guided by the flat and oblique oriented surface of the articular processes (Fig.1.23). Flexion and extension stretches the fibre if the disc in between the vertebrae as the vertebrae tilts a bit as they move. The flat surface and oblique orientation of the articular processes is a hinder for pure rotation or lateral bending in the same time as it guides the motion. When the vertebra bodies moves, the disc in between are stretched on the posterior side (flexion) or the anterior side (extension).

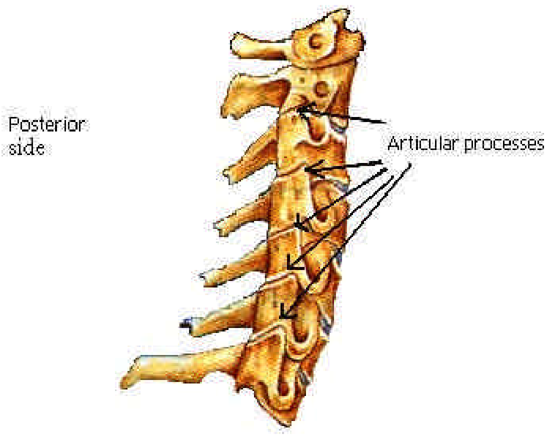


Fig. 1.23 Lateral view of C1-C7 and T1, the oblique position of the articular processes is visible. (Adapted from Marieb, 1998)

The movement of the head is controlled by the overall movements of each of the cervical vertebrae added together. However, certain vertebrae have unique types and range of motion. The atlas allows slight forward and a larger backward head nodding due to its joint with the occiput. The joint between axis and C_3 lets slight rotation of the head relative to the neck.

The lower cervical spine has a particular behaviour coupled movement, replicated mechanically in the designed dummy, due to the spatial orientation of the facet joints.
The neck moves with a coupled motion (both rotational and lateral bending in the same time) as two neighbouring vertebrae rotates around the axis perpenicular to the articular processes. This coupled motion is a main characteristic for the biomechanics of the neck. For example, during lateral bending to the left, the left articular facet glide downward and the right one glide upward (Fig.1.24).

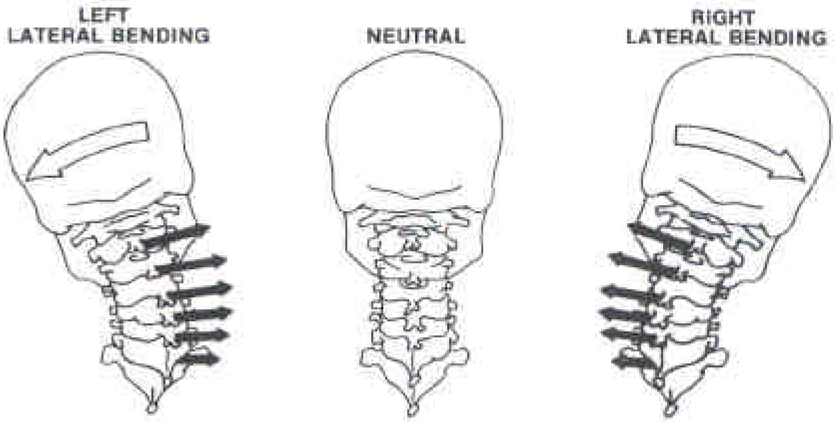


Fig. 1.24 Spine orientation due to lateral bending (adapted from Panjabi[26])

Figure 1.25 shows a model for the neck, by Kapandij. The model is a mechanical solution for the coupled motion in the lower part of the neck, pure rotation between C_1-C_2 and no rotation between C_0-$C_{1.}$ However, the model does not enable flexion and extension in any coupling in the lower part of the mode. Even if the coupled motion always occur in the lower part of the cervical spine, it is still possible to turn the head to one side and hold the horizontally. Pure motion of this type can be done since a compensated motion takes place in the upper cervical spine, with C_1 and C_2 that has a different shape as mentioned before.

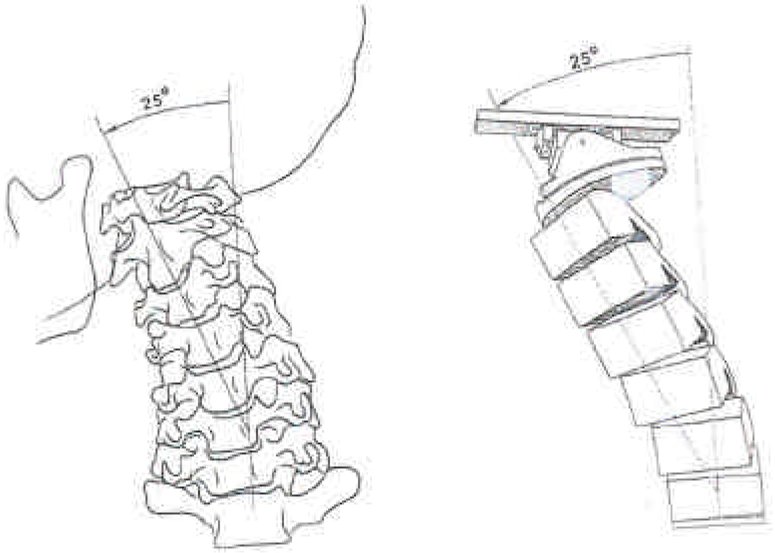


Fig. 1.25 To the left picture, vertebrae in the cervical spine during rotation. The rotation also results in the lateral bending. To the right, a model to illustrate the coupled motion in the cervical spine. (Adapted from Kapandji 1978)

1.4 RANGE OF THE MOTION

The range of the motion in the cervical spine is depends on several factors and differs from person to person. Age and individual elasticity influence on the flexibility in the spine and determines the maximal range of the motion for the neck. White and Panjabi presented representative values for the ranges of rotation of the cervical joints based on various studies (Table 1.2). In most of these studies, measurements were derided from radiographic examination of volunteers and, therefore, represent in vivo ROM. The load magnitudes that produces the motions are not known. The atlanto-axial joint (C_1-C_2) allows for much axial rotation, due to the unique shapes of the vertebrae and articulations: roughly half of the axial rotation of the neck occurs at this joint. The occipito-atlantal joint (C_0-C_1) allows for much flexion-extension and only little axial rotation. Both C_0-C_1 and C_1-C_2 allow for little lateral bending compared with the lower joints. C_7-T_1, at the transition of the cervical and the thoracic spine, allows for the least rotation in all directions.

Coupling	Flexion-Extension (total)	Lateral bending (one side)	Rotation (one side)
Occiput-C_1	25°	5°	5°
C_1-C_2	20°	5°	40°
C_2-C_3	10°	10°	3°
C_3-C_4	15°	11°	7°
C_4-C_5	20°	11°	7°
C_5-C_6	20°	8°	7°
C_6-C_7	17°	7°	6°
C_7-T_1	9°	4°	2°

Table 1.2 Range of motion for flexion-extension, lateral bending and rotation. (White and Panjabi, 1990)

In Table 1.3, Kapandji´s values for the range for motion is presented. These values were found using oblique radiographs in the extreme positions (for flexion and extension). They differ from Panjabi and White's values in the range for extension-flexion between C_2-C_7, rotational motion between occiput-C_2 and C_2-C_7. The differences can be due to the selected samples and measurement method.

Coupling	Flexion-Extension (total)	Lateral bending (one side)	Rotation (one side)
Occiput-C_2	20°-30°	8°	24°
C_2-C_7	100°-110°	45°	80°-90°

Table 1.3 Range of motion in the cervical spine according to Kapandji.

1.5 LOAD-DISPLACEMENT BEHAVIOUR

Mechanical characteristics of biological tissues are necessary in determining the dummy design properties. These properties vary on every individual as it happens with the ROM depending on same factor such as age, gender, etc. The stiffness attributes of the neck in main movements should be considered to get a reliable dummy. Motion segments or ligaments like neck ones typically have a nonlinear, sigmoidal shape load-displacement curve (Fig.1.26).

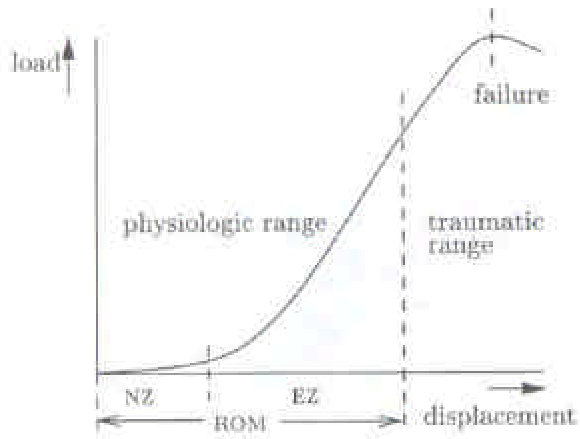


Fig. 1.26 Typical Load-Displacement curve for biomechanical structures (adapted from de Jager)

The curve is divided into a physiologic and traumatic range. The physiologic range starts with the neutral zone (NZ) in which little load is needed to deform the structure. The transition from the neutral to the elastic zone (EZ) is characterized by a substantial increase in load. Throughout the elastic zone, the curve is usually fairly linear. The name “elastic zone” reflects that, on release of the load, the specimen will return to the state it had before being loaded. The physiologic range of the motion (ROM) is the sum of the neutral and elastic zone, and represents the total amount of displacement that the biomechanical structure can sustain without being damaged (the design ODD-neckII replicates this particular human behaviour thanks to the use of some rubber cushions between adjacent vertebrae).

Because it is difficult to identify precisely when damage (micro trauma) starts and the ROM ends. The traumatic range starts when the micro trauma occurs within the structure and ends with failure of the structure , which is characterized as a substantial drop in load. Load and displacement at failure define the failure strength of the structure.

For motion segments, the NZ represent the region where ligaments are lax such that small loads produce large displacement. In the EZ, the ligaments are stretched and their resistance increases, resulting in increased motion segment stiffness.

In most experimental studies on motion segments, loads are applied statically: load is increased in small increments after which the specimen is given time to reflex before the displacements are measured. Consequently, viscous effects are minimized and elastic characteristic are obtained. In few cases, loads are applied quasi-statically or dynamically: load is continuously increased at a relatively low, respectively, high rate, and viscoelastic characteristic are measured.

Stiffness and flexibility are used to describe the load-displacement curves. Stiffness is defined as the ratio of the load applied to the displacement produced (stiffness is defined also as the slope of the tangent to the curve at a certain load or displacement). Flexibility is the reciprocal of stiffness. Due to the nonlinear behaviour of the curve the stiffness varies along it, therefore different methods were developed to estimate this slope, so the range of load and displacement for which the stiffness were calculated should be given.

1.6 BIO-MECHANICAL MOTION DURING AN IMPACT

When the body is subjected to an impact, like the one in a car crash, it is exposed to forces of greater magnitude than it can obstruct. Muscles react instinctively with a counteraction but the head and neck will move in a non-physiological motion, the muscles are not strong enough to control the motion. It is not yet clear how the neck muscle reflexes influence on the cervical spine during an impact but they respond in time to affect the motion.

In a rear end car collision the struck car is accelerated. This means that the car occupant is pushed forward by the seat back. If the seat is equipped with a head-rest there is usually space between the skull and the head-rest. This means that the skull, due to its inertia, tends to lag behind when the trunk is accelerated forward. An extension motion of the neck will follow. This motion is abruptly interrupted either when the head is reached by the heat-restraint or when the head and neck reach the maximum extension angle (Figure 1.27). The motion scenario described above is in the literature called whiplash motion.

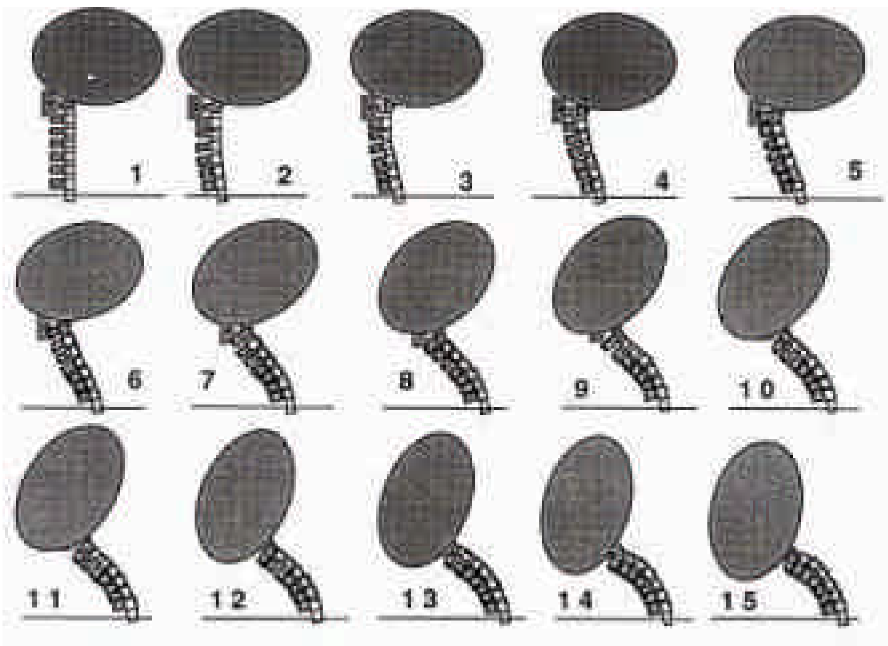


Fig. 1.27 Schematic representation of whiplash motion (Svensson, 1989).

The common whiplash extension-flexion motion can be divided in three phases (Figure 1.28).

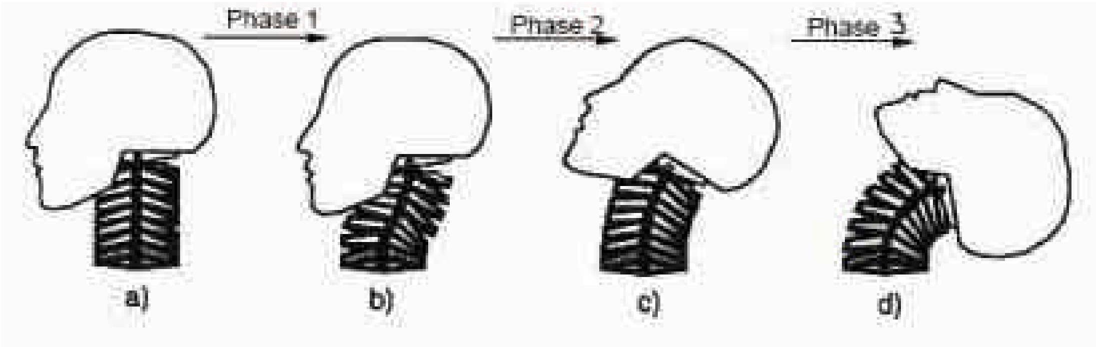


Fig. 1.28 The three phases of a whiplash motion: 1) Linear rearward motion from the initial posture; 2) The head starts rotating rearward; 3) Rearward rotation until fully extended position.

The first phase lasts about 100 ms. The seatback pushed the torso forward while the head stays still. This can be considered as a linear rearward motion of the head relative to the torso. There is no angular motion of the head occurring. Hence, the spine adopts an S-shape. This motion causes the non physiologic motion of the neck, called retraction. During retraction, the upper part of the cervical spine goes into flexion and the lower one into extension. When both parts reach their limits for maximum flexion respective extension the linear rearward motion will stop and the head starts rotating rearward. Some theories are based on that whiplash injuries take place during the retraction. It causes pressure changes in the spinal canal which can lead to soft tissue damages.

At the end of the Phase 1 the linear rearward motion of the head is abruptly decelerated at the same time as the head starts rotating backwards. This is explained by the fact that the upper cervical spine reaches its limit for maximum flexion while the lower cervical spine reaches the limit for full extension. In the second phase (from 100 ms to 150 ms), the extension motion of the upper cervical spine still accelerates, but the lower part of the cervical spine goes into a less extended position.

In the third phase (from 150ms to 250ms), the whole cervical spine goes into full extension until it is stopped by the structures of the neck. Another theory of the cause of whiplash is failure of the fact capsules (especially if the head was rotated in the time of impact) during extreme extension in this phase of the crash.

Following this extension motion, the whole cervical spine returns towards its initial posture by means of the elastic energy that is stored in the neck structures at the fully extended posture. This flexion part is normally much less violent than flexion motions that are seen in frontal collisions and it is much lower than the initial extension motion.

In frontal and side impacts, the neck usually experiences the same type of inertial loading from the head as in rear end collisions(Fig. 1.29). During the initial phase of these neck loading situations, the head normally undergoes a horizontal translational displacement relative to the torso. This is particularly evident in frontal and rear end collisions. This translational motion is called protraction for forward motion and retraction for rearward motion. The neck is exposed to very significant mechanical loads when the end of the normal range of protraction or retraction of the neck is reached, and neck injuries may well occur at this point. This may be one explanation for the fact that modern head restraints do not provide better protection. They may simply come into play too late, after the neck has exceeded the maximum range of retraction motion and gone into extension.

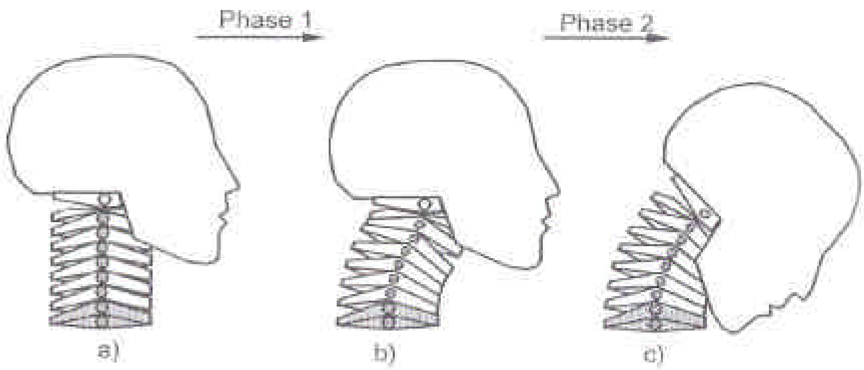


Fig. 1.29 Schematic drawing of the head-neck motion during a frontal collision.
Phase 1: Protraction motion
Phase 2: Flexion motion

1.6.1 The symptoms of injury on the muscles

A rear- end collision begins a sequence of events affecting the cervical spine, its joints, ligaments, and musculature. A similar sequence can be evoked in any injury causing abrupt hyperextension or hyper flexion of the neck. The impact abruptly propels the body in a linear horizontal direction. Due to inertia, the head remains in its initial position then abruptly moves in the opposite direction. The movement of the head must be in the direction of flexion or extension of the cervical spine.

The abrupt movement of the neck occurs before the neck muscles can relax and permit this motion. Acting upon the unprepared muscles initiates an acute stretch reflex. In a rear-end impact the head moves abruptly backward causing acute hyperextension of the cervical spine causing an acute stretch reflex of the neck flexors (hyperextension causes an overstretch and inappropriate contraction of the neck flexors with residual flexor disability). A head-on collision would cause the opposite reaction with the stretch reflex being of the extensor muscles as the head proceeds forward and the neck acutely flexes.

If the stretch is acute, abrupt, and overwhelming, the muscle fibrils sustain an injury. This injury is to the intrafusal fibers essentially and the extrafusal fibers (contractile fibers) if the forces is appropriately severe. This initiates a proportionately strong stretch reflex with a forceful muscular contraction. The injury is to the fibrils and not to gross muscle bulk, thus there is usually no gross haemorrhage or edema. As major nerves are not initially involved, there may be no immediate pain, hyperesthesia, parasthesia, or a paresis.

There is microscopic edema and hemorrhage combined with fibril injury. This edema and hemmorrhage organize and form a myofascial fibrositic nodule. This “traumatic fibrositis” remains as a focus of irritability causing more muscle irritability with muscular contracture, fibrous contracture, emotional reaction, and thus chronic pain and limitation..

Irritability or “spasm” is claimed and noted by patients and examiners. Initially the muscles may be temporarily “paralized” which accounts for so many patients the morning after an accident stating their inability to “lift their heads off the pillow”.

When the muscles become overwhelmed and they are elongated past their physiological limits the fascial connective tissue, the tendons, the ligaments and the articular capsules are also overstretched and thus injured. As the joints are caused to exceed their physiological limits they essentially sublux causing all periarticular tissue damage of a subluxation. The magnitude of the reflex muscular contraction in response to the stretch is proportional to the abruptness and the strength of the stretching force.

1.6.2 General symptoms of injury

The injury symptoms have been described in large number of papers (e.g.Sturzenegger et al., 1995; Spitzer et al., 1995). The symptoms of injury following neck trauma in rear-end collisions include pain, weakness or abnormal responses in the parts of the body (mainly the neck, shoulders and upper back) that are connected to the central nervous system via the cervical nerve-roots. Vision disorder, dizziness, headaches, unconsciousness, and neurological symptoms in the upper extremities are other symptoms that have been reported (Deans et al., 1987; Hildingsson, 1991; Nygren et al., 1985; Spitzer et al., 1995; Sturzenegger et al., 1995; Watkinson et al., 1991). The neck injury symptoms appear to be very similar for all impact directions (Minton et al., 2000). It is important to distinguish between initial symptoms and long term symptoms (Krafft, 2000). Long term (chronic) whiplash symptoms appear to be associated with central pain sensitisation (Sheather- Reid and Cohen, 1998; Johansen et al., 1999). The exact origin of this pain sensitisation has not been established. Successful treatment methods could possibly provide a clue. Byrn et al. (1993) reported significantly reduced symptoms during a time period after sub-cutanious sterile water injections on the back of the neck. Bogduk (2000) reported pain relief in about 50 percent of the patients after coagulation of the small nerves that innervate the facet joint that is associated with the painful dermatome.

Soft tissue injuries have been found in several different structures and locations in the neck region in experimental studies and autopsy studies. In a recent study Yoganandan et al. (2000) reported injuries to several ligaments, the intervertebral discs and the facet joint structures. Siegmund and Brault (2000) and Brault et al. (2000) presented indications of muscle injury due to eccentric muscle loading in the early phase of the neck motion in rear impacts. Taylor et al. (1998) reported interstitial haemorrhage in cervical dorsal root ganglia in an autopsy study of victims who had sustained severe inertial neck loading during impacts to the torso or to the head. The structures around the ganglia were mostly uninjured. These findings correlate to experimental findings in pigs of nerve cell membrane dysfunction in cervical spinal root ganglia reported by Svensson et al. (2000).

The risk of neck injury to rear seat occupants was only about 50% of the risk of neck injury for front seat occupants in rear-end collisions (Kihlberg, 1969; States et al., 1972; Carlsson et al., 1985; Lövsund et al., 1988; Otremski et al., 1989). The injury symptoms following neck trauma in rear-end collisions include pain, weakness or abnormal response in the neck, shoulders and upper back as well as vision disorders, dizziness, headaches, unconsciousness, and neurological symptoms in the upper (States et al., 1972; Nygren et al., 1985; Hildingsson, 1991; Watkinson et al., 1991; Spitzer et al., 1995). Spangfort (1985) used Figure 1.30 to describe the stages of the symptoms. Findings similar to those of Spangfort (1985) were reported by Deans et al. (1987).

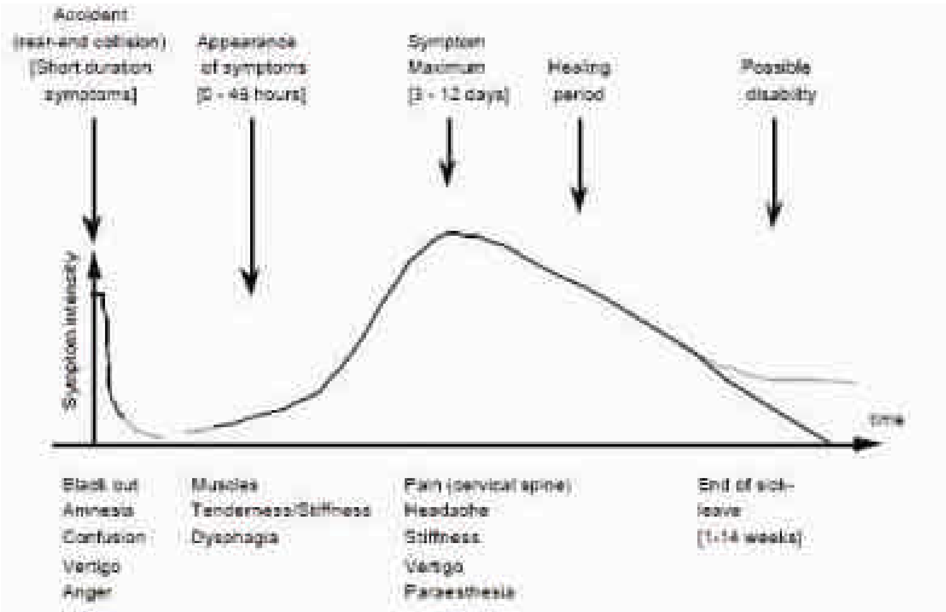


Fig. 1.30 The stages of the neck injury symptoms sustained in a rear end collision (Spangfort, 1985).

2. EXISTING MULTI-DIRECTIONAL DUMMY NECKS

Several crash test dummies have been developed in the last years in order to replicate the human behaviour and improve the safety in the automotive field. Depending on the sort of impacts to study, different characteristic are implemented in the dummy.
For car seat evaluations or ratings there are two basic methods available: a static evaluation and a dynamic evaluation. The static measurement estimates the quality of the head restraint position of which ratings are documented by RCAR (Research Council for Automobile Repairs), while the dynamic method relates the quality of the entire seat to measurements in rear impact dummies. Evaluation of a crash test dummy is done by comparison between test results from the dummy and human subjects from an impact. The human subjects are either volunteers or PMHS. Tests done with volunteers are usually done with speeds around 6-9 km/h, to eliminate the risk of causing injuries to the volunteers. No speed limit has to be taken in concern while using PMHS but the effect from the neck muscles is lost.
The evaluation of biofidelity has its main focus on kinematic behaviour and does not specifically focus on loads measured in dummies as compared to those calculated in human testing. The reason is that there are assumptions made for the calculations of the human neck loads, which are not easily comparable to dummy load measurements. For instance, the Occipital Condyle joint in a dummy is a hinge joint, while in the human the OC is surrounded by other load bearing tissue

2.1 BRIEF DESCRIPTION

Zero-Order Head-Neck Model and Advanced Head-Neck Model

In early 1980, Some head-neck models were made in the effort to resemble the human as much as possible. Examples of such models are the Zero-Order Head–Neck Model (Kabo and Goldsmith, 1983) and the Advanced Head-Neck Model (Winters and Goldsmith, 1983). These models had a head of a cadaver (water-filled), manufactured vertebrae, muscles substitutes (all muscles of the neck, except some of the smaller hyoid muscles, were modelled) and ligament elements. The models made a comparison between head motion and muscle behaviour possible. The simplified and robust models described below replaced these very advanced models.

Hybrid III

This is the most commonly used dummy for both frontal and rear impacts, although it was originally designed for frontal impact. It was developed for General Motors.
This dummy neck has a flexible base component (butyl elastomer) and three vertebrae substitutes in form of rigid aluminium washers. A steel cable runs through the centre of the neck, to give axial strength (Fig. 2.1). There are a range of Hybrid III dummies depending on the population that represent; the most widely used is 50th percentile which represents an average size and weight of the men population.
It is able to move slightly in the lateral direction but since it is not validated in this direction it is usually not used in these crash tests. During a frontal or rear-end test with Hybrid III, the movement of the neck differs from human behaviour. The major difference is the absence of the S-shape.

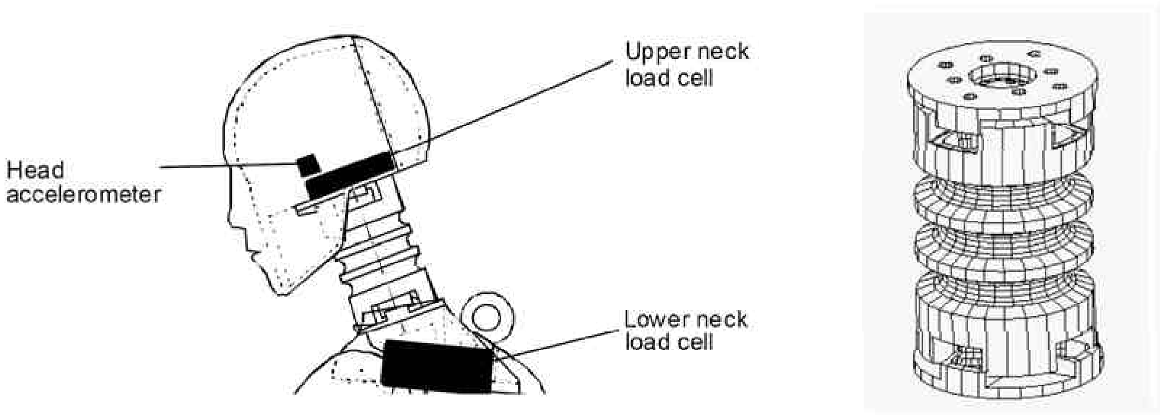
Fig. 2.1 Hybrid III 50th percentile Head&Neck (adapted from Denton Atd, Inc.)

The performance at high severity rear impact seems rather good (Prasad, 1997), but the performance in low severity (whiplash) cases is rather poor (Scott 1993, Davidsson 2000, Cappon 2001a). The main problems in low severity impact relate to the rigidity of the spine, the limited flexibility of the hip joints and the stiff neck, showing no head lag. Even the addition of a more flexible 2D TRID neck (Thunnissen, 1996), does not result in a biofidelic response.
Seemann et al. (1986) found the Hybrid III neck far too stiff to respond in a humanlike manner in the sagittal plane. Deng (1989) reported that results from a mathematical model of the Hybrid III neck indicated that the neck has a torque response similar to that of the human neck but has a higher shear response. Foret- Bruno et al. (1991) compared the Hybrid III dummy with a cadaver in simulated rear-end impact using a headrest closely fitted to the head, to minimise the relative movement between head and torso. The cadaver showed no sign of injury. In spite of this, very large shear forces at occipital level were registered in the Hybrid III test. The authors concluded that the human head can be moved relative to the torso with no stresses in the neck, but this is not the case for the dummy. In volunteer tests, McConnell et al. (1993) found that during the acceleration phase of a rear-impact, when the occupants body was pressed against the seat-back, the spinal curvature straightened. This in turn caused an upward motion of the head and thus an elevated head contact point on the head-restraint. In a comparative study using volunteers and a Hybrid III dummy, Scott et al. (1993) found that the dummy was less prone to ramp up along the seat-back than were the volunteers.

SID, BIOSID, EUROSID, WORLDSID

They were designed for a deep study of lateral impacts. The SID was the first attempt to study the side impacts developed by National Highway Traffic Safety Administration (NHTSA). BioSID is based on the Hybrid III neck so it doesn’t have any relevant modifications in the neck.

EuroSID (Fig. 2.2) has been created by European Experimental Vehicles Committee (EECV).There are two versions of this dummy called EuroSID 1 and EuroSID 2. Both of them have a neck made by a composition of metal discs and rubber elements with special joints to head and chest to allow a realistic motion of the head relative to the chest.



EuroSID 2 **WorldSID**

Fig. 2.2 Neck schematics from EuroSID and WorldSID(adapted from EEVC and ISO respectively)

Recently, International Standard Organization (ISO) has completed the development of the WorldSID dummy (Fig. 2.2) with the cooperation from several countries in the world, including Europe and USA joining their efforts. The neck of the WorldSID has a central deformable element, bracket to adjust pre-impact orientation and a shroud to avoid unrealistic interactions.

All the dummies described previously are designed to represent 50th percentile of average-size men.

Thor

This 50th percentile male dummy was developed by National Highway Traffic Safety Administration (NHTSA). It is a frontal impact dummy with a more biofidelic frontal response than the Hybrid III, which has been evaluated for rear impacts as well, but it has also been used in lateral and oblique test.
It has a multi-directional neck to enable accurate head motion. The neck (Fig. 2.3) is made from a series of aluminium discs and rubber pucks which are bonded together and also has compression springs attached to simulate the effects of the musculature. The neck gets the S-shape in frontal impact tests. This dummy neck is an improvement compared to the Hybrid III neck.

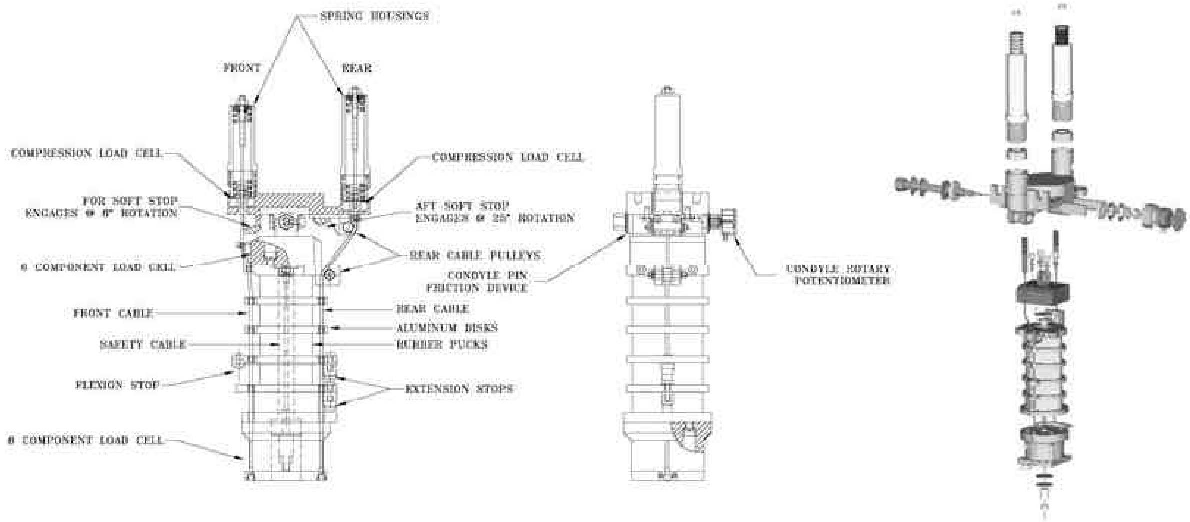


Fig. 2.3 Thor neck (adapted from NHTSA)

RID2 dummy

The RID2 prototype dummy was originally designed and built within the European Whiplash Project (Cappon, 2001b). The dummy was later updated to a commercial version, called RID2, by FTSS (Cappon, 2001a). The RID2 is a 2½ D dummy, which means that it is not meant for 3D use, but yet can handle oblique rear impacts. The back shape of the dummy is based on the UMTRI data and reflects the 50th percentile male human back shape, ensuring human like seating. The dummy was evaluated against low severity volunteer tests (5g) and higher severity PMHS tests (12g). Most of the responses are biofidelic and the RID2 shows the typical s-shape in the neck. However, the dummy showed limited ramping up and lower neck rotations. Furthermore, the dummy was found to be repeatable and reproducible.

BioRID II

This dummy was designed by Chalmers University of Technology (Davidsson, 1999). The BioRID II was developed as a 50th percentile male dummy to measure responses in low speed rear-end impact tests. The BioRID I was updated to the prototype BioRID P3, which is being manufactured by R.A. Denton Inc. under the name BioRID II. The dummy has a multi-segment spine, representing all the vertebrae in the human body. The dummy shows biofidelic behaviour in most responses, compared to volunteer experiments performed earlier at a low delta V (7-9 km/h). Also the typical s-shape in the neck, causing head lag, is present in the BioRID II. The only differences found relate to the spine straightening and the head rotation, but these are rather minor. The dummy was shown to be very repeatable, but also sensitive to ringing of the spinal structure (Kim, 2001).

Svensson and Lövsund (1992) developed and validated a Rear Impact Dummy-neck (RID-neck) that can be used on the Hybrid III dummy. It was designed to resemble the human anatomy to enable a trajectory, and angular range of motion similar to that of the human in the sagittal plane.

The BioRID-neck (Fig. 2.4) consists of seven cervical vertebrae, which are connected by pin joints similar to door hinges. This specific design allows producing the S-shape observed

in human necks during rear-end collisions, which is important to develop the injuries caused by traumatic neck motion.
The vertebrae are made of acetal plastic. Two cables are running through the segments along the cervical spine on the posterior side. They replicate the superficial muscle groups and create an improved relation between the retraction and extension components of the neck motion. Between the vertebrae there are rubber blocs fixed, which can be exchanged in order to be able to control the resistance to the motion.
The range of motion of the neck was increased in comparison to the values found in volunteer experiments, to allow some kind of hyperextension and -flexion. At present the BioRID-neck allows only flexion-extension motions. However the BioRID may help to find out more about how seatbacks, head restraints, and other vehicle characteristics influence the likelihood of whiplash injury.

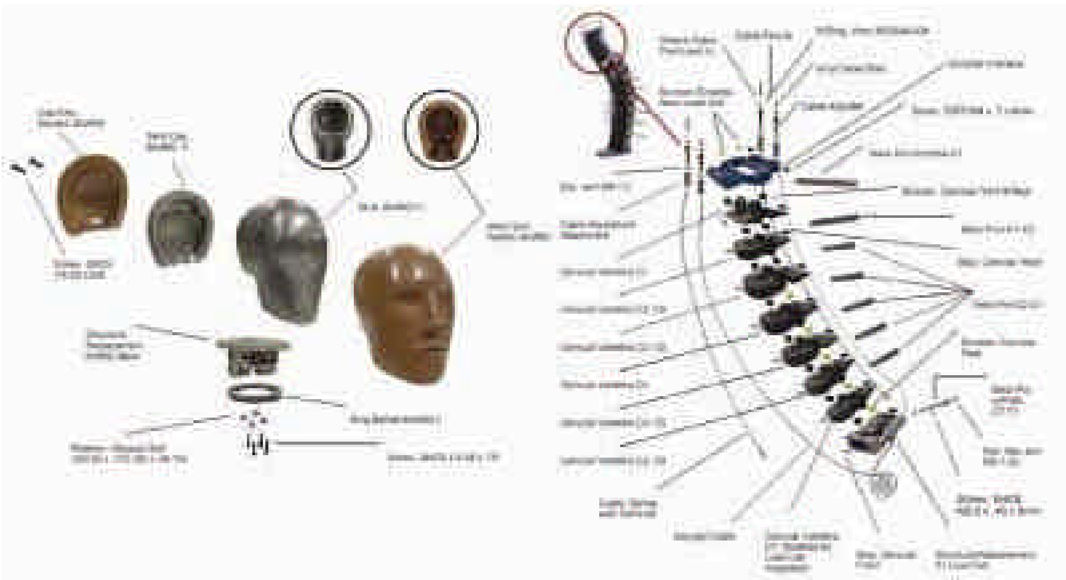


Fig. 2.4 BioRID II Head&Neck (adapted from Denton Atd, Inc.)

2.2 REVIEW ODD-NECK II

The chapter of the skeletal body aims to present the basic components of the dummy neck (vertebrae and joints) and chapter of the intervertebral stiffness properties aims to present the mechanical characteristics of the dummy neck in the previous work (for more details to see [9]) .

2.2.1 ODD-neck II : Skeletal body

The dummy neck is inspired by BioRID, this neck stand out because of its resemblance to the human body and it pretends to imitate the human behaviour. It is comprises seven vertebrae and one occiput interface (Fig 2.5). The upper cervical of the prototype consist of three vertebrae: an occiput interface (C_0), C_1 and C_2 and several joints to connect them.

Fig. 2.5 General 3D model of the skeletal part of the neck with all the vertebrae and joints

Upper cervical. The neck attaches to the head by means of the $C_{0,}$ it has a circular shape with an inside hole across it, a horizontal pin joint is used to link the C_0 to the head. A thin plastic layer is placed between C_0 and the head interface to avoid movements from the occiput and fix its position (Fig. 2.6).

Fig. 2.6 Occiput interface, C0, views: 3D model, wireframe

The C_1 vertebrae (Fig.2.7) has a particular profile rather different than any other vertebrae to allow the dummy to replicate the particular behaviour of the upper cervical spine. It has an underneath threaded hole to fasten perpendicularly to C2 while the upper part consists of two jutting out parts, which are round off in the top to let the piece rotate around the joint. A horizontal hole goes across this underneath parts to allow the connection with C0.

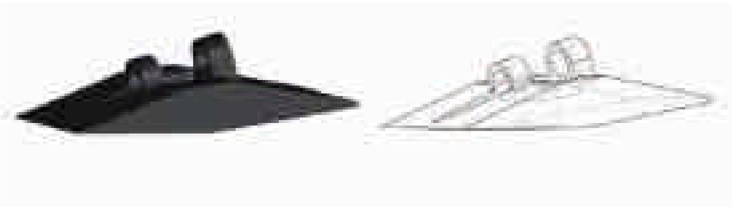


Fig. 2.7 C1 views: 3D model, wireframe

The mechanical joint solution between occiput and C1 is a box with two pin-joints. It has round sides next to the pins to allow the rotation of the box across them. Each of the pin joints permits one movement (Fig.2.8), flexion and extension in the upper joint and lateral bending in the lower one. The box is placed without an angle related, so pure moments are obtained in both directions. Based on review of literature [25], axial rotation was neglected between these vertebrae to simplify the design.

Fig. 2.8 Box views: 3D model, manufacture prototype and possible movements between C0C1: flexion/extension and lateral bending

The joint between C_1 and C_2 is made of two main parts, a tube part which together with a pin part connect the vertebrae to C_2 allowing flexion and extension and a thread weld to the tube that screws down to the C_1 (Fig. 2.9). Between C_1 and C_2 only flexion, extension and pure rotation are possible. Based on review of literature [25], lateral bending was neglected between these vertebrae to simplify the design, always within a reasonable humanlike behaviour.

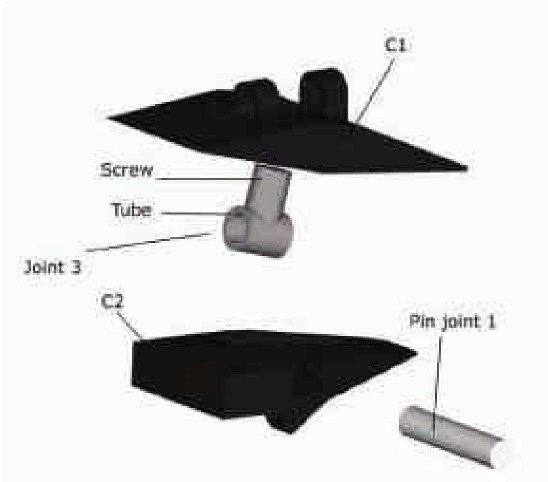


Fig 2.9 Joint solutions between C1 and C2

Lower cervical. The lower cervical comprises vertebrae from C3 to C7 and unlike the human body, where they have different size, they are identical in the dummy (Fig. 2.10). It is based on BioRID but it has slope sides to enable the lateral bending/axial rotation

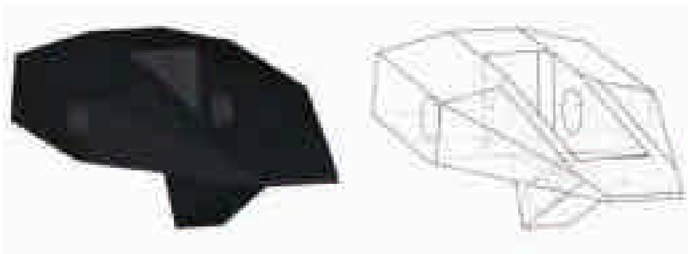


Fig. 2.10 Lower cervical vertebra views: 3D model, wireframe

The main characteristic of the lower cervical spine is the coupled movement produced between lateral bending and axial rotation (Fig. 2.11). This is fairly achieved thanks to the joint solution previously explained in the upper cervical. The screw threaded part of the joint will substitute the articular processes providing guidance for flexion-extension and hindering lateral movement. Due to the connection between the tube and the screw, the rotation around the tube of the joint allows flexion-extension between vertebrae in the lower neck. The upper vertebra is screwed down to the joint and rotates around it. Due to the angle, which represents the average slope of the articular processes in the neck, a coupled movement of lateral bending and rotation similar to human body is created.

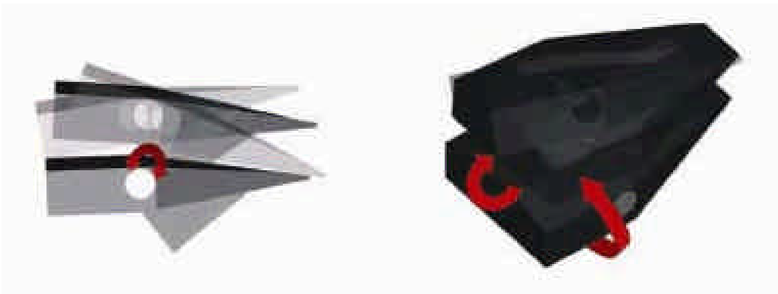


Fig. 2.11 C3-C4. Possible movements: flexion/extension and couple lateral bending/axial rotation

Neutral position. The human cervical spine has a curvature of about 37° from C_0 to C_7 in the default position so the vertebrae are placed in the neck with an average angle of 5.3° to each other. The neck curvature follows the same curvature as the BioRID cervical spine. At this default position all the cervical joints centres are placed at an imaginary arch with a radius of 190mm and a sector of 37º. To achieve the neck spine curvature the intervertebral rubber bumpers fit in the space between vertebrae at their particular position (Fig.2.12).

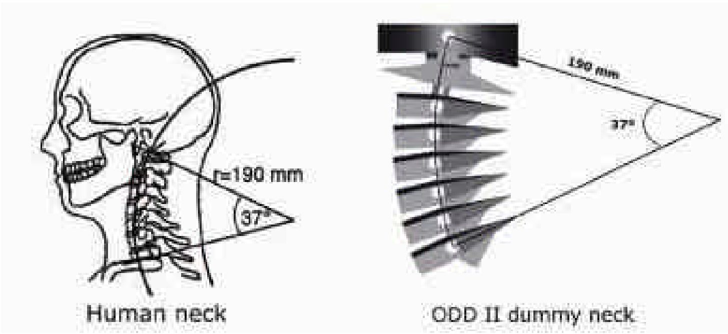


Fig 2.12 Default cervical spine curvature

Range of the motion. The neck is designed to perform a range of motion (Table 2.1) within human values. Without any limitation place between vertebrae, the range of motion was designed based on in vivo studies [26] derived from examination of volunteers and including muscle effect of the neck. The movement is controlled by the shape of vertebrae and joints. Some assumptions have been made for simplicity and the range of motion is the same in the lower cervical for each vertebrae based on the mean value of the available data. The range of motion was increased a bit on every value for two reasons: on one hand to leave enough space between vertebrae to place rubber that may replicate stiffness properties of the neck; on the other hand to allow a movement a bit over the in vivo data because its values are based on volunteer test and the neck itself could go a bit further.

Motion		C_0-C_1	C_1-C_2	C_2-C_3
Flexion-extension	In vivo	25	20	16.4
	ODD-II	27.8	37.1	27.5
Lateral bending (one side)	In vivo	5	5	9.4
	ODD-II	17	-	17.18
Axial rotation (one side)	In vivo	5	40	6
	ODD-II	-	180	10.74

Table 2.1 Range of motion of the prototype without any limitation between vertebrae and in-vivo values

2.2.2 ODD-neck II: intervertebral stiffness properties

In the mechanical behaviour of the human cervical spine, the stiffness has a important role. From all available options to get the stiffness neck properties, two mechanical solutions were developed to get the right one: place rubber between the vertebrae, which compression will restrict the motion and the cables to attach two vertebrae that will limit the range of motion by stretching it.
The cushion of the rubber was made with two layer of polyurethane, fist layer imitates the stiffness of the neutral zone while another harder polymer replicates the range of the motion stiffness. It was made from cast polyurethane because of its good physical properties such as durability, good elasticity and wide range of hardness (stiffness) available.

Lower cervical. The rubber piece for flexion has a 9x13mm^2 area with 4,5mm of thickness which are made by 2,5mm layer 40 Shore A and another one of 2mm 20 Shore A. For extension a 10x14mm^2 rubber piece of 10mm thickness composed by a 6mm 55 Shore A and 4mm 20 Shore A layers was used (Fig. 2.13).

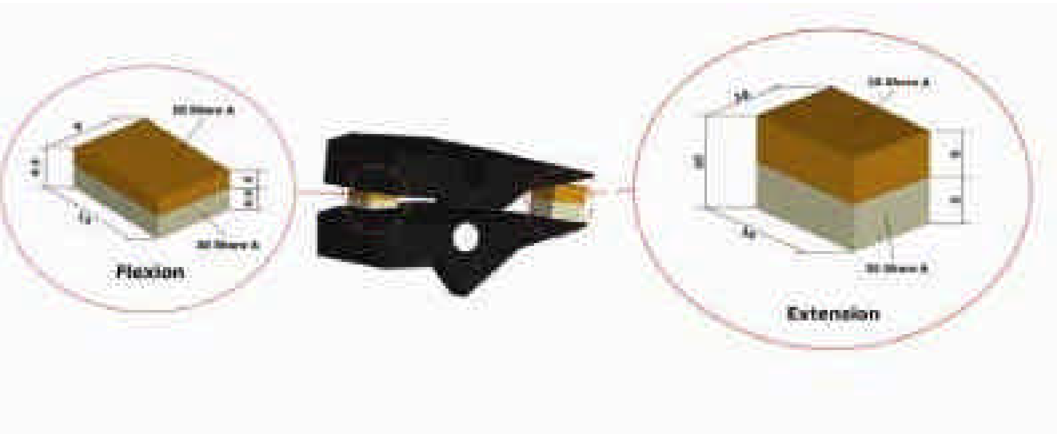


Fig. 2.13 Rubber dimensions placed between vertebrae in the lower cervical

The results of the tests of the behaviour of the rubber can see on the graphic resemble the human values (it was chosen load-displacement curves of the lower cervical spine of the human behaviour from studies of Camacho) with its particularities of hardening the stiffness as the load is increasing (Fig. 2.14).

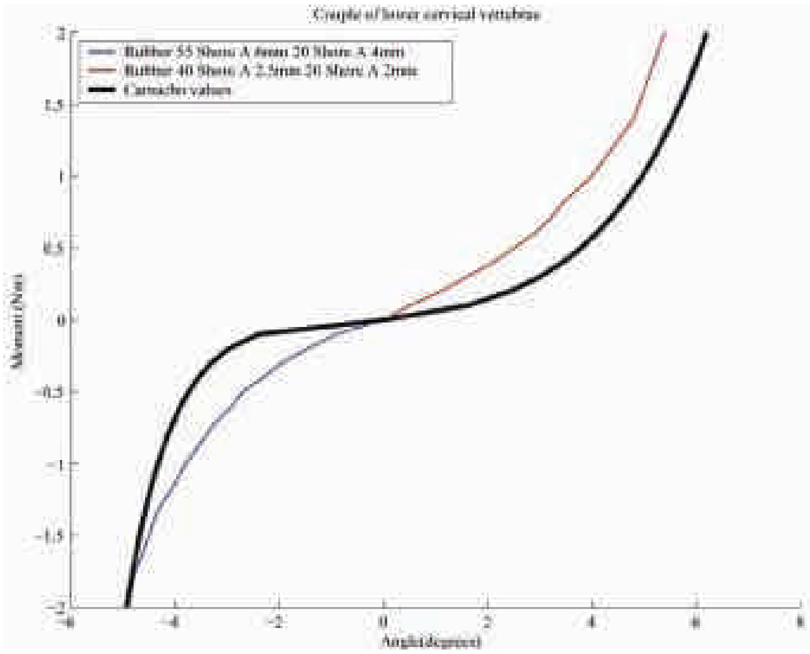


Fig. 2.14 Final model values of the dummy neck rubbers in flexion/extension

The couple patterns in the lower cervical spine are one of its most important characteristics. On this part of the cervical spine lateral bending movement is always coupled with axial rotation. The dummy neck replicates this behaviour, every movement in lateral bending has an axial rotation associated.
A rubber was placed between vertebrae to get the stiffness in lateral bending and axial rotation. It was selected a trapezoid shape (Fig.2.15), that fits perfect in the interverterbral space on the sides and has a smooth limiting the movements.

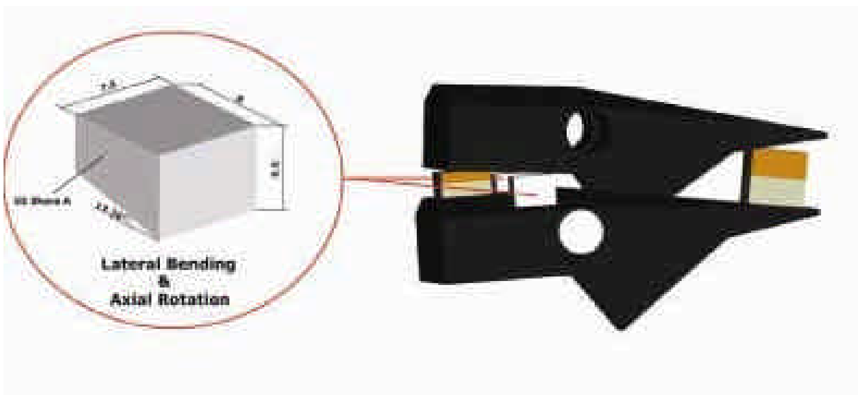


Fig. 2.15 Rubber cushions to limit lateral bending and axial rotation of the vertebrae

The final rubber pieces placed on the dummy are 55 Shore A hardness with 7.5mm thickness the trapezoid (8mm base, 5.5mm height, 13,26º angle). The tests were performed measuring the lateral bending (Fig. 2.16) response and it was converted afterwards to get the axial rotation.

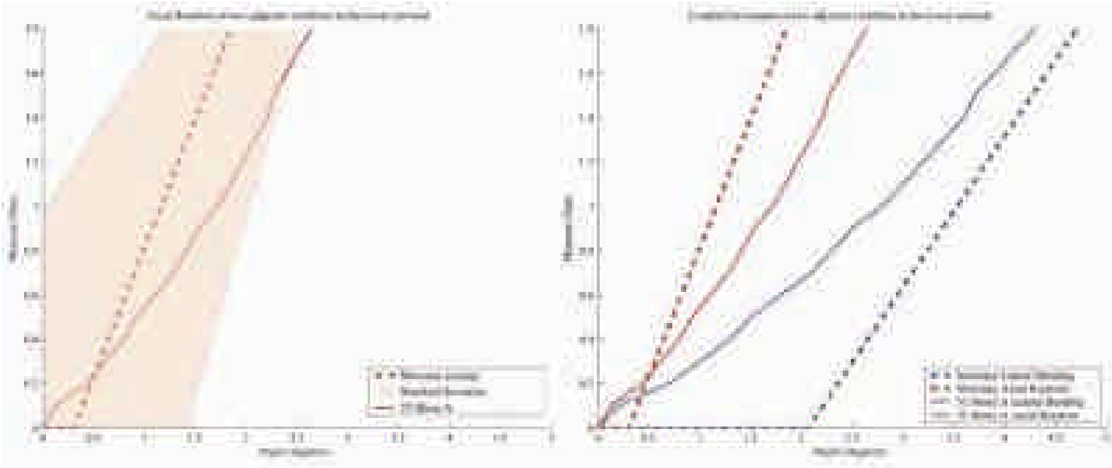


Fig 2.16 Test results: coupled movement in the lower cervical spine

Upper cervical. The data from Camacho were used in upper cervical to define the intervertebral properties of the dummy too.

Rubber pieces were placed between C_0 and C_1, but in these two vertebrae unlike lower cervical the rubber is glued to upper vertebrae instead of the lower one because of the C_1's shape. The final rubber piece for extension is a two layer polymer of 13mm thickness and 14x11.5mm^2 area with a 9mm layer of 19 Shore A hardness and a 4mm layer of 55 Shore A. A 9mm thickness and 19x13 mm^2 25 Shore A was chosen for flexion (Fig.2.17).

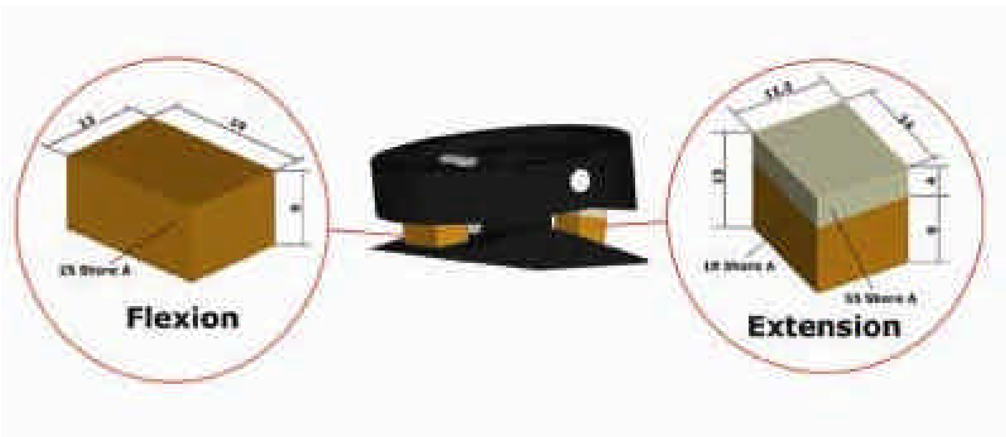


Fig. 2.17 Rubber cushions placed between occiput interface and C1 for flexion/extension

The tests were extended to higher loads up to 2.0 N·m showing a really close behaviour to the human values (Fig.2.18).

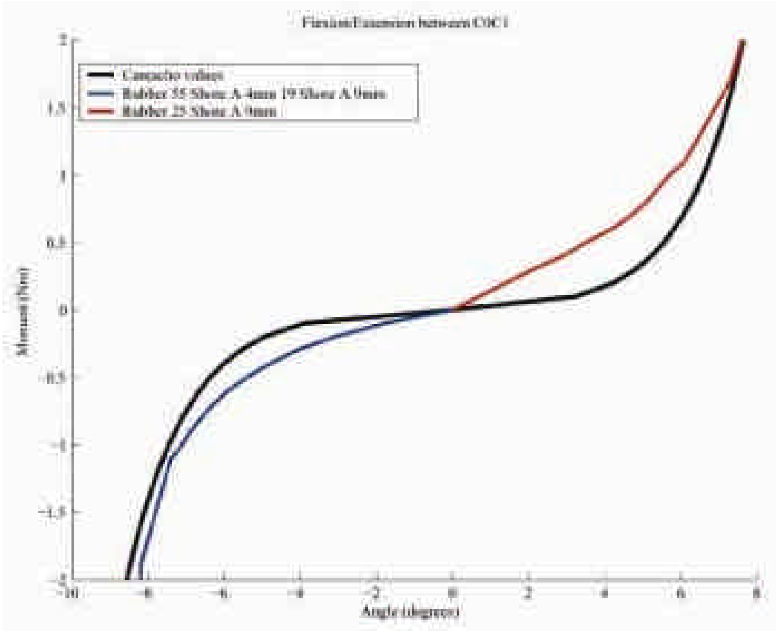


Fig 2.18 Final values of the dummy neck rubbers in flexion (positive)/extension (negative) between C0C1 together with Camacho values

The rubber cushions were glued on the lower vertebra, C2. In order to not to interfere with the axial rotation movement between these two vertebrae that will be described later, a thin layer of plastic film, like the one used for cooking, is placed on top of the rubber to avoid the friction produced between the polymer and C1 (Fig.2.19).

Fig. 2.19 Rubber bumpers located between C1 and C2 vertebrae of the dummy

The flexion's one is a two layer polymer 10mm thickness and 14x10mm2 with a 6mm 60 Shore A layer and 4mm 23 Shore A hardness while the extension is 7.5mm thickness 15x9.5mm with one sheet 3mm 55 Shore A and 4.5mm 19 shore A.

The tests show that both rubbers replicate quite closely in vitro values(Fig.2.20).

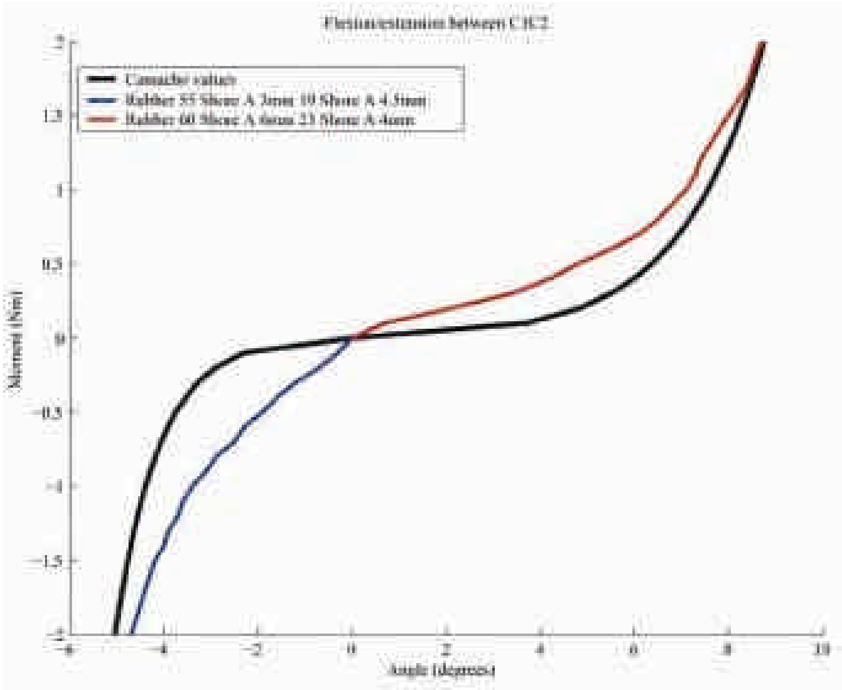


Fig. 2.20 Final values of the dummy neck rubbers in flexion (positive)/extension (negative) between C1C2 together with Camacho values

In order to simplify the design of the neck the lateral bending between C_1-C_2 was neglected, therefore it just appears between C_0-C_1 by means of the box that connect them. Two rubber pieces were glued on each side of C1. Both of them are identical 10mm thickness 9x14mm2 made by a polymer 65 Shore A hardness with the following behaviour (Fig.2.21):

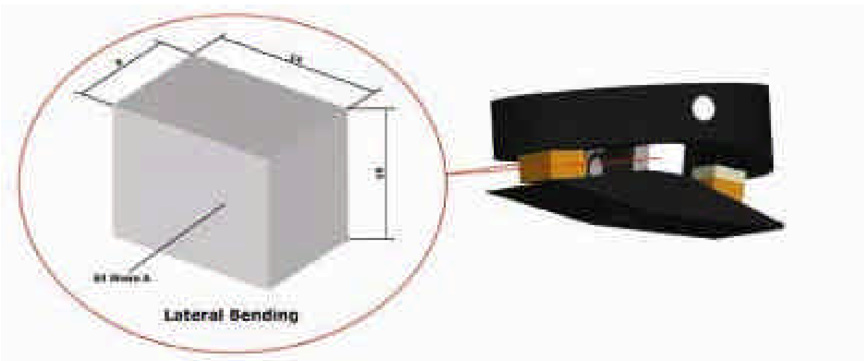


Fig. 2.21 Rubber pieces placed between the modified load cell and C1

The results of the test with the rubber placed between vertebrae together with the data from Panjabi and Oda are plotted (Fig. 2.22):

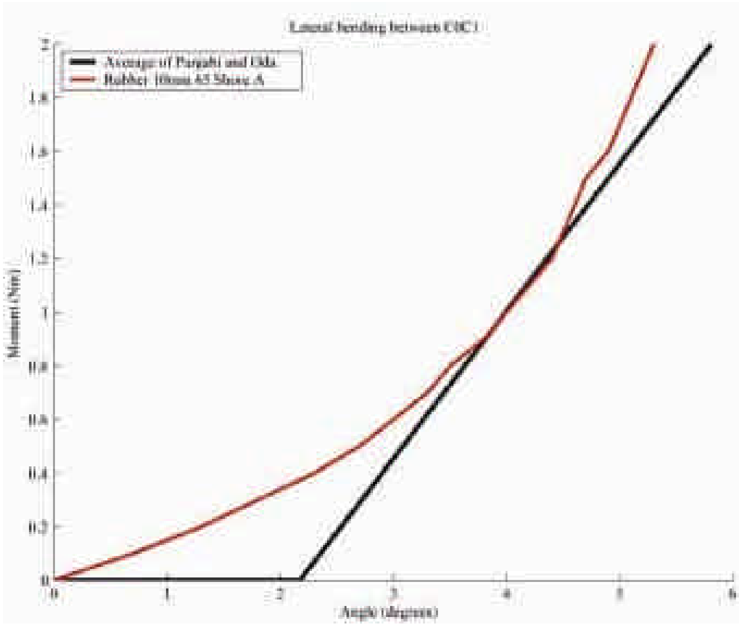


Fig. 2..22 Final values of the rubbers in lateral bending between C0C1 together with an average of Panjabi and Oda

The axial rotation between C_0-C_1 was neglected in the upper cervical spine to make the design of the dummy easier, it just appears between C_1-C_2. A system based on a cable and its stiffness properties was the solution to restrict axial rotation in the upper cervical spine. The rope should be placed somehow that delimits the movement independently from the flexion-extension between vertebrae. It is fixed in one side to the pin joint between the vertebrae and in the other side to C1 (Fig. 2.23).

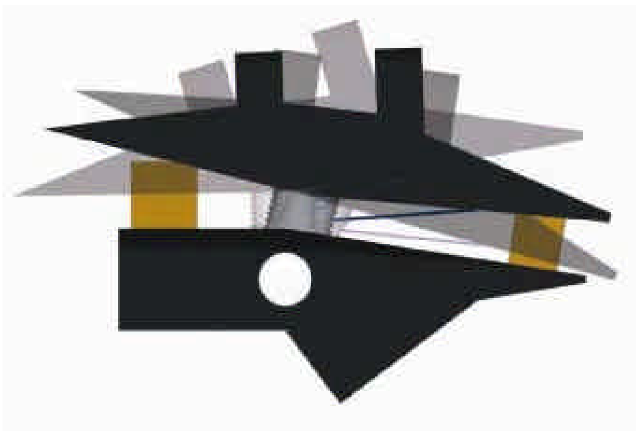


Fig. 2.23 Position of the cable between vertebrae, its elongation is not affected by flexion/extension movement

Although some flexion-extension movement takes place between vertebrae, with this particular location (link to the pin joint and to C1) the tensile behaviour of the cable remains identical for every possible position because it moves together C1 around the pin joint of the lower vertebra.

The axial rotation between C1C2 is constricted by the elongation of the cable. When C1 rotates around the pin joint, the cable should stretch because the distance between the two fixing points increases from the initial value (Fig. 2.24).

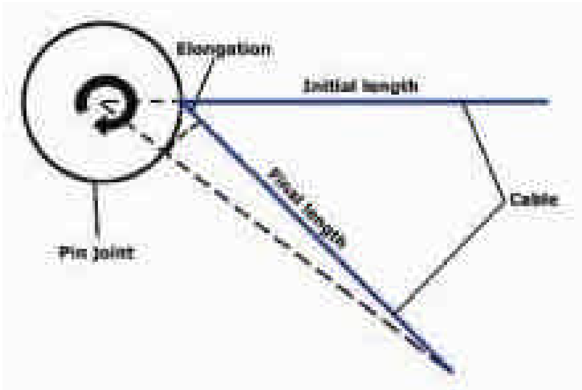


Fig. 2.24 Elongation of the cable due to the applied moment between C1C2

The chosen material for the cable has been nylon, because its elasticity is closed to the require values.
The values that agree better with the desired stiffness properties were found out with Ø0.45mm nylon cable that should be 21.56mm length while the distance between the two fixing points of the nylon is 21.09mm. It should be noted that the nylon doesn't have a perfect elastic behaviour so a small permanent deformation comes out. After the tests were performed, the increase of the distance from the initial value of 500mm was measured. A 0.1% of permanent elongation that defines the deviation of the results was found out. It is possible to predict the theoretical behaviour of the cable with a length of 21.56mm in the vertebrae in terms of Moment-Angle (Fig. 2.25).

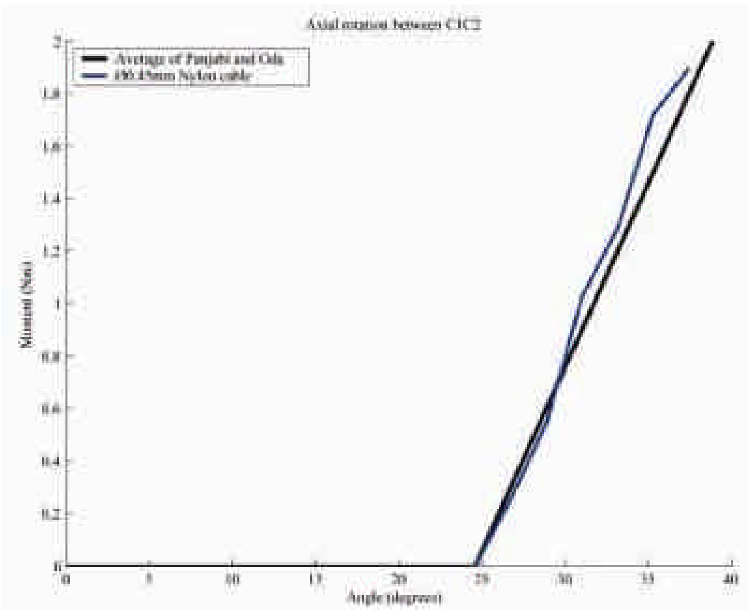


Fig. 2.25 Theoretical values of the cable in axial rotation between C1C2 together with an average of Panjabi and Oda

Due to the small length of the cable, it is really difficult to get an accuracy of 21.5660mm; therefore a system for adjusting the tension of the cable in the vertebrae was the chosen alternative method (Fig.2.26) used to define the stiffness properties in axial rotation between C1C2. On one side the cable is fixed to the lower part of C1; three holes were drilled: two in the bottom surface of the vertebra and a thread one on the side. Then the cable goes through the two bottom holes and it is fixed by means of a screw that presses it, being able to adjust the tension of the nylon. On the other side, it is fixed to the screwed part of the pin joint that connects C1 and C2. A hole with two different internal characteristics is drilled through the entire pin joint. On one side it is thread to allow inserting a screw that has in turn, a hole in the middle to let the cable go through it. By the screw movement it is possible to adjust the cable length.

Fig. 2.26 Methods used in the dummy neck to adjust the tension of the cable between C1C2

2.2.2 Conclusions

ODD-neck II, this new three dimensional dummy neck not only does it replicate the movement pattern, furthermore, it has a biofidelic performance that closely simulates the behaviour of the human neck both range of motion and stiffness. A main improvements is that it would be applied in lateral, oblique, roll-over (where the vehicle is vaulting), far side (on the opposite side of the dummy), frontal and rear-end impact. The muscle response was neglected so a simulation of the human neck muscle system could be implemented between vertebrae.
The final aspect of the global neck is (Fig.2.27):

Fig 2.27 Global OmniDirectionalDummy neck II: manufacture prototype and 3D model

The static response of the dummy neck in the main four movements of he neck is shown below, the neck fits into the head of the BioRID by means of attaching the occiput to the neck load (Fig.2.28).

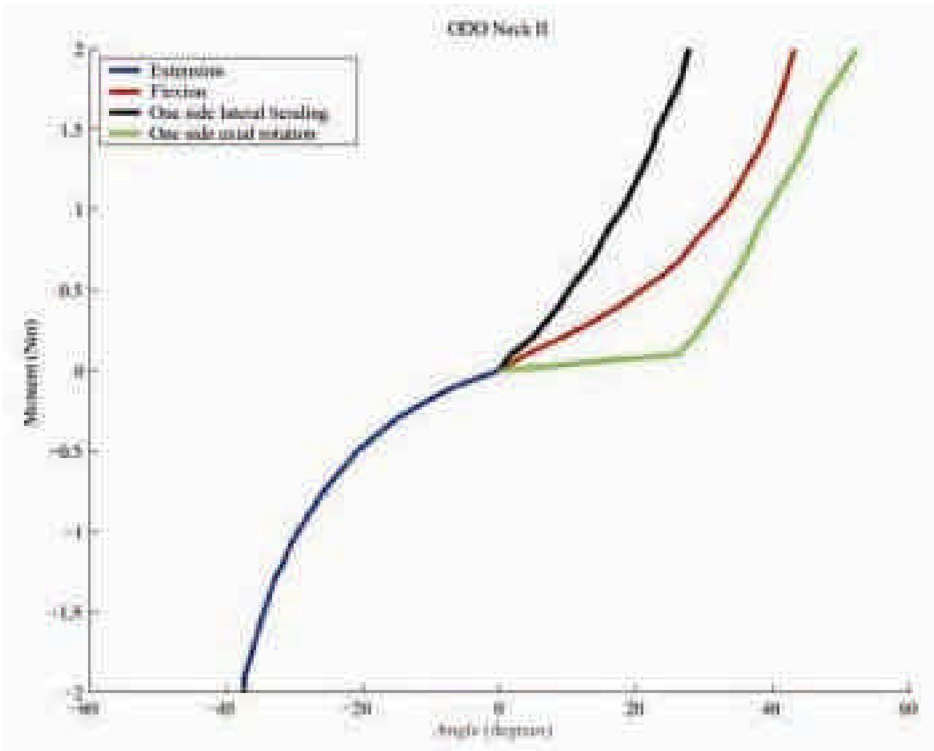


Fig. 2.28 Static response of the ODD II dummy neck in the main four movements of the neck

3. REVIEW OF THE NECK MUSCLE SYSTEM

Skeletal muscles generate and exert forces. Thus, they are the basic element of the mechanics of the human movement (Herzog, 1994). In a car collision, skeletal muscles cannot fully control the movements and position of the human body since torques due to external forces acting on the human body joints are likely to greatly exceed those generated by muscle. Under such conditions muscles can act as natural restraint system that resists an externally imposed movement and load. Thus, it seems to be reasonable to hypothesize that, when the human body is subjected to inertia and contact forces resulting from a car collision, skeletal muscles might still have considerable effects on its responses. This hypothesis can be supported by the results of experimental studies that suggested or indicated differences in the kinematics and kinetics of the head-neck complex between post-mortem human subjects (PMHS) and volunteers (Wismans et al., 1987; Thunnissen te al., 1995), as well as between volunteers with relaxed and tensed muscles (Mertz and Patrick, 1967; Mertz and Patrick, 1971; Ono te al., 1997). However, a detailed experimental investigation of the effects exerted by muscles on the responses of the human body under transient loads is greatly limited for both ethical and technical reasons; method for in-situ measurement of the muscle forces are not well established (Herzog, 1994). Therefore, methods for mathematical modelling, such as multibody and finite element, might be an alternative way to conduct such an investigation. The available models of muscles and the musculoskeletal system were developed to analyze primarily the movements of the human body when fully controlled by nervous system and muscles. When the human body segments are accelerated to 15-70 g in ms, as in a typical car crash, the role of muscle is significantly different that of daily activity in four important ways:

- skeletal muscles account for 40-45% of the mass of the human body. They are not rigidly attached to the bones. When the human body segments are rapidly accelerated, inertial forces cause relative movements of the bones and muscles. These movements affect the kinematics of the human body segments.
- Muscles, after excitation by the nervous system, are capable of generating force, i.e. developing moment about the joints. This moment lies in the 100-515 Nm range (Tracy, 1990). When muscles are not activated, they behave as passive structures that resist external load.
- Muscles can absorb and dissipate a certain amount of energy (Armstrong and Waters, 1968). The stretched can store about 9000 J/kg of elastic energy (Shorten, 1993).
- The phenomena of impact conditions occurs rapidly. The relation between the process of generating muscular force and time is important. The control function of the nervous system and human perception should not be overlooked.

When there is contact between an impactor and the human body, for instance between a car bumper and a pedestrian, the relative movements of bones and muscles, and energy absorption by muscles, are the prominent factors. When there is no contact between the impactor and the human body, the muscles can be regarded as structures that limit the motion of the human body segments. For instance, forces generated by muscles can influence the movements of the neck and head in a frontal car crash (Bowman and Robbins, 1972; Wismans et al., 1987).

3.1 TYPE OF MUSCLE CONTRACTION

Joints make a skeleton potentially movable, and bones provides a basic system of levers, but bones and joints cannot move by themselves. The driving force, the power behind movement, is muscle tissue.

The basic physiological property of the muscle tissue is contractility, the ability to contract, or shorten. In addition, muscle tissue has three other important physiological properties. Excitability (or irritability) is the capacity to receive and respond to a stimulus, extensibility is

the ability to stretch, and elasticity is the ability to return to its original shape after being stretched or contracted.
These four properties of muscle tissue are related, and all involve movement.

Several types of muscle contraction have been identified, including twitch, isotonic, isometric, titanic, concentric and eccentric.
Twitch. A momentary, spasmodic contraction of a muscle fiber in response to a single stimulus (such as an electric current or a direct stimulation of a motor neuron) is a twitch. It is the simplest type of recordable muscle contraction. The short, jerky action of a twitch is usually produced artificially in order to record the response on a graph called myogram (Fig 3.1)

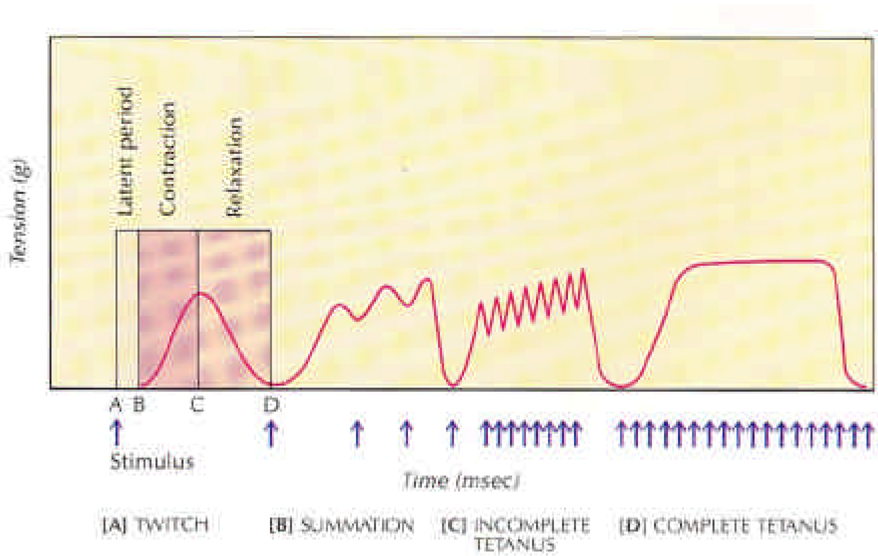


Fig. 3.1 Myograms of several types of contraction . (A) Twitch. The muscle is stimulated once at point A. There is a short latent period (A-B), only a few milliseconds, before the contraction actually begins at point B and peaks at point C. The contraction lasts about 0,04 seconds. The following relaxation period (C-D) lasts about 0,05 second.
(B) Summation. Summation occurs when more than one stimulus is given before the muscle has relaxed completely after first stimulus.
(C) Incomplete tetanus. Incomplete tetanic contraction when a muscle is just beginning to relax when another stimulus is received.
(D) Complete tetanus. Complete titanic contraction occurs when the rest periods between stimulations become so short that the muscle cannot relax at all.

Tetanic contraction. If a muscle receives repeated stimuli at a rapid rate, it cannot relax completely between contractions. The tension achieved under such conditions is greater than the tension of a single muscle twitch, and is called summation of twitches (Fig 3.1.B). A more or less continuous contraction of the muscle is called a titanic contraction or tetanus. When incomplete relaxations are still evident between contractions, the muscle is said to be in a state of incomplete tetanus (Fig.3.1.C). Complete tetanus occurs when the muscle is in a steady state of contraction, with no relaxation at all between stimuli (Fig 3.1.D).

Isotonic and isometric contractions. When a muscle contracts by becoming shorter and thicker, the contraction is called isotonic because the amount of force, or tension remains constant as movement takes place (Fig. 3.2 A)
In contrast, if the load on a muscle is greater than the tension developed by muscle , the muscle retains its original length, and the contraction is isometric (Fig. 3.2 B).

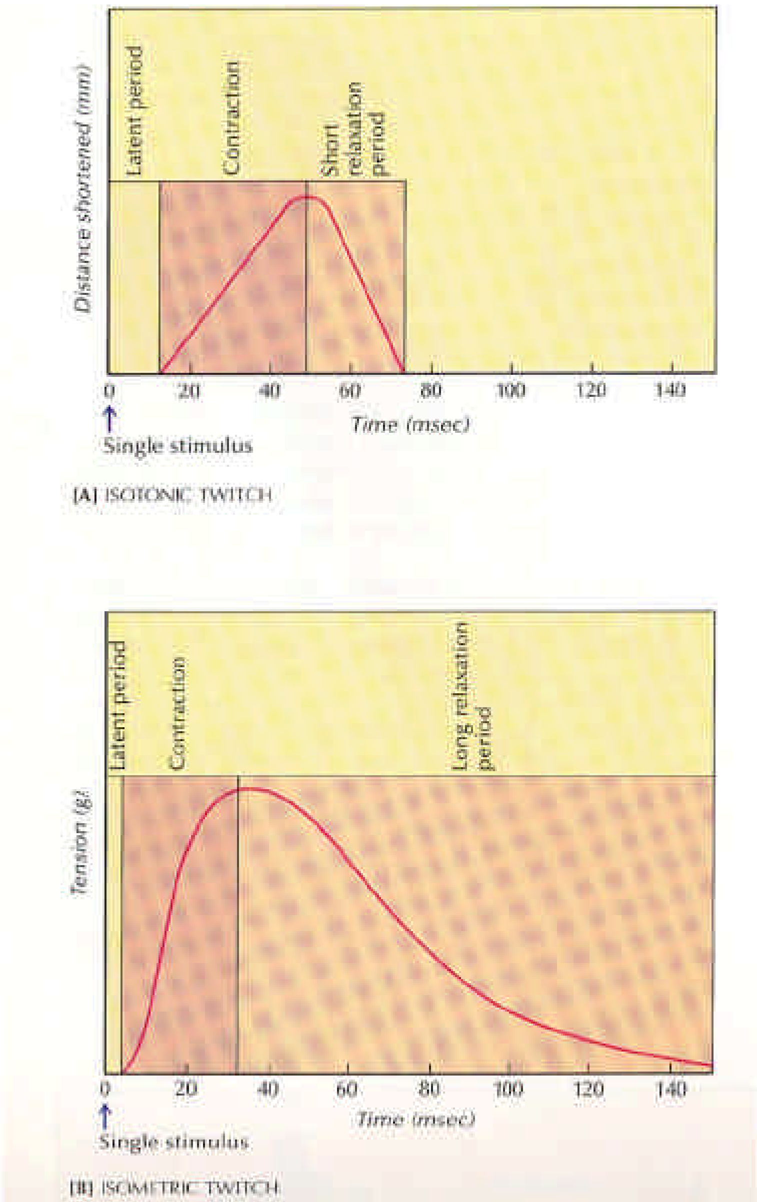


Fig. 3.2 Comparison of an isotonic twitch and isometric twtitch

Concentric contractions. When a muscle is activated and required to lift a load which is less than maximum titanic tension it can generate, the muscle begins to shorten. Contractions that permit the muscle to shorten are referred to as concentric contractions. The force generated by muscle is always less than the muscle's maximum. As the load the muscle is required to lift decreases, contraction velocity increases.

Eccentric contractions. During normal activity, muscles are often active while they are lengthening. As the load on the muscle increases, it finally reaches a point where the external force on the muscle is greater than the force that the muscle can generate. Thus even though the muscle may be fully activated, it is forced to lengthen due to the high external load. This is referred to as an eccentric contraction. There are two main features to note regarding eccentric contractions. First, the absolute tensions achieved are very high relative to the muscle's maximum titanic tension generating capacity. Second, the absolute tension is relatively independent of lengthening velocity.

3.2 REVIEW OF THE BASIC MECHANICAL PROPERTIES OF THE SKELETAL MUSCLE

From an engineering point of view a muscle can be treated as an actuator controlled by the nervous system. Its force can be divided in two components: passive and active ones.

Passive muscle. The passive force is developed when the inactivated muscle is either slowly stretched beyond its resting length (i.e., spring-type behaviour) or subjected to rapid length changes (i.e., damper-type behaviour) figure 3.3.

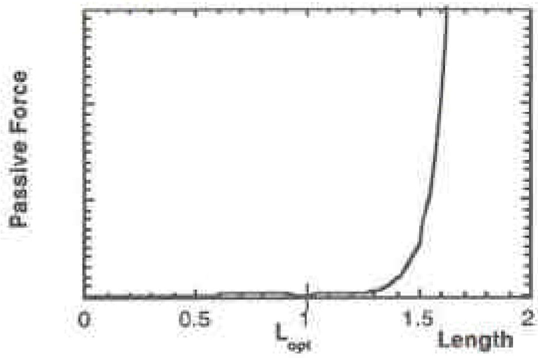


Fig. 3.3 Schematic representation of the passive isometric muscle force-length relationship

Most of the data for the material properties of muscles was obtained with the assumption that muscle carry loads in only one dimension, i.e. tension and compression. In tension, skeletal muscles exhibit strongly nonlinear behaviour (Yamada, 1970; Keir et al., 1993). The stress-strain curve has a "toe" region at low strain level. However, at high elongation, muscle stiffness increases rapidly.
The ultimate stress and strain depend on age; the greatest elongation of muscle tissue has been observed in the 10-19 age group (Yamada, 1970). Mathematically, the relation between force and elongation in inactivated muscles is usually described by means of an exponential function (Hatze, 1981). The stress in inactivated muscles depends also on strain rate (McElhaney, 1996; Bahler et al., 1968; Grive and Armstrong, 1987).
Briefly stated, the following two aspects of behaviour of inactivated muscles are the most important. The first is that muscles exhibit nonlinear viscoelastic properties, i.e under high elongation, Young's modulus increases rapidly. The relation between muscle stress and high strain rate is not well described in the literature. The second aspect is that material constants vary with age and experimental conditions, consequently any results that are available in the literature should be interpreted carefully.

Active muscle. For the purpose of analyzing the role of muscles in impacts, the muscles can be considered as force generators with complex, nonlinear characteristic. The active force is generated after excitation of the muscle by nervous system. The force generated by a certain muscle depends on some features of the muscle itself (for instance morphology and physiological cross section), the neural excitation, the extension, and the velocity of shortening or lengthening. This force is typically described by the force-length, force- deformation velocity, and force-time relationships, e.g., Bahler et al. (1968), Pierrynowki and Morrison (1985), Pitman and Peterson (1989), Winters (1990), and Herzog, (1994) figure 3.4.

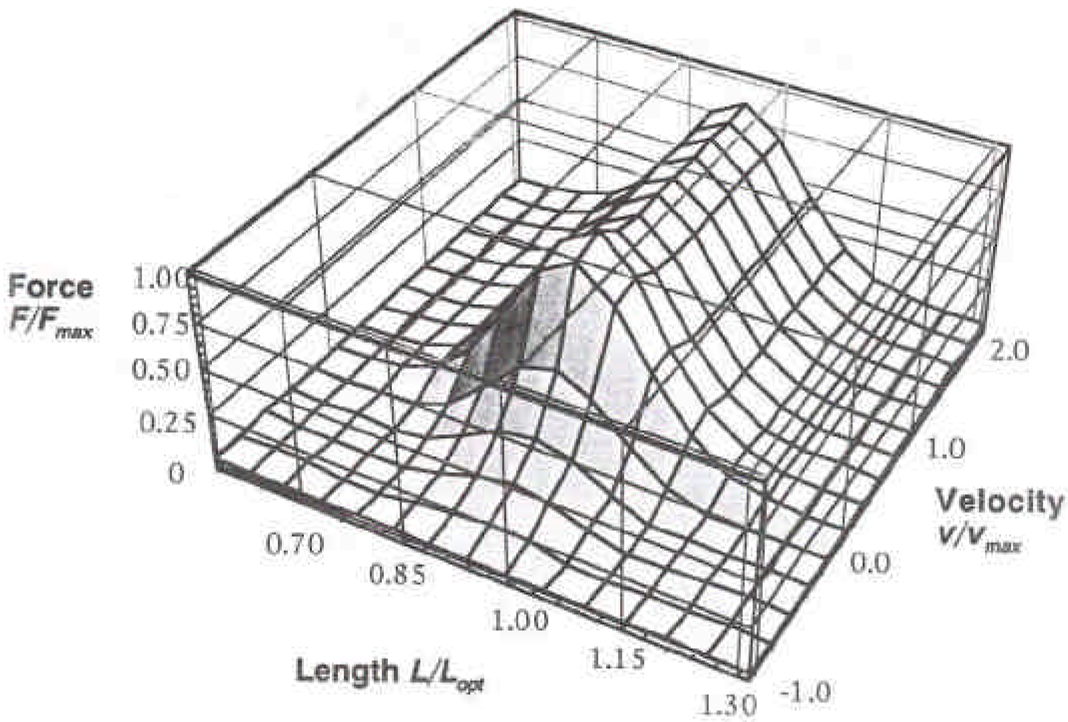


Fig. 3.4 Muscle force as a function of the muscle length and shortening-lengthening velocity. F_{max} is the maximum isometric force, l_o is the length at $F_{max,}$ and v_{max} is the maximum velicity of shortening. Negative velocities correspond to shortening. Results were obtained by means of a muscle model developed by Winters and Stark.

The active force-length relationship F_L describes force-length properties of the maximally activated muscle under isometric conditions. It exhibits maximum force at optimum muscle length L_{opt} (Figure 3.5). When the muscle is either shorter or longer than L_{opt}, its F_L decreases.

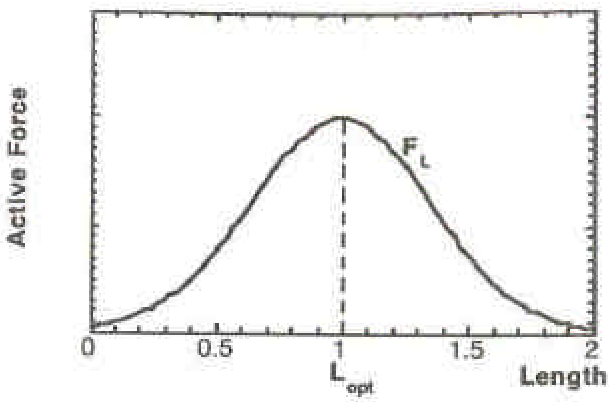


Fig. 3.5 Schematic representation of the active isometric muscle force-length relationship

The general behaviour of the force-length relationship (i.e., the sum of the active and passive forces) varies between muscles, which merely arises from differences in amount and distribution of the connective tissue (Carslon and Wilkie, 1974). For muscles containing a large amount of the connective tissue, this relationship is a continuously increasing function (Figure 3.6).

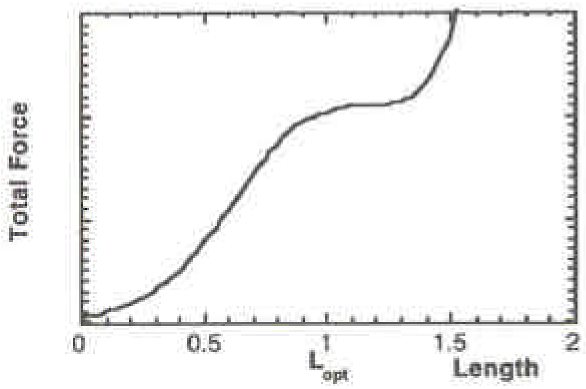


Fig. 3.6 Schematic representation of the total isometric muscle force-length relationship (i.e., sum of the passive and active muscle forces) for a muscle having a large amount of the connective tissue.

On the other hand, for muscles having a relatively low amount of the connective tissue, the force-length relationship initially decreases according to F_L during a stretch beyond their L_{opt}. Then, the total force increases due to a rise in the passive force-length relationship (Figure 3.7). This type of muscle force-length relationship has been observed in human rectus femoris (Herzog and Keurs, 1988) and gastrocnemius muscle (Zuubier et al., 1994).

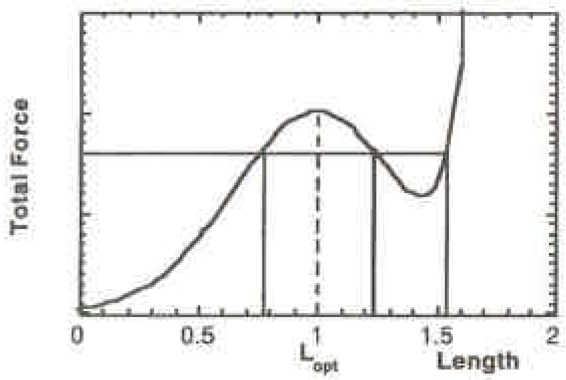


Fig. 3.7 Schematic representation of the total isometric muscle force-length relationship (i.e., sum of the passive and active muscle forces) for a muscle having a low amount of the connective tissue.

This suggests that human skeletal muscles may operate on the descending limb of the force-length curve within a physiological range of joint motion. For this reason, in the mathematical modelling literature it is typically assumed that the total force-length relationship of the human cervical muscles exhibits a minimum of around 0.4-0.5 of the maximum isometric force at lengths beyond L_{opt}, e.g., De Jager (1996) and Van der Horst et al. (1997). An important feature of this type of force-length relationship is that it implies that up to three distinct muscle lengths can correspond to a given value of the muscle force.

The force-deformation velocity relationship describes the relationship between the force of the maximally activated muscle and the instantaneous rate of change in the muscle length. The force-deformation velocity is typically obtained at the optimal muscle length. When the velocity of muscle deformation is zero, the muscle contracts isometrically and its force is determined by the active force-length relationship F_L summarized in Figure 3.5. When the muscle shortens (concentric contraction), its force decreases to zero at the maximum shortening velocity v_{max} (Figure 3.8). A classical description of the relationship between the muscle force and its shortening velocity has been formulate by Hill (1938,1970) and is referred to as the Hill hyperbola. For large lengthening velocities, the force generated by muscle yields an asymptote. This asymptote had been estimated in the literature to be in range of 1.25 to 1.8 of the isometric force (Pierrynosnwski and Morrison, 1985; McMahon, 1987). Thus, from a point of view of mathematical modelling of a skeletal muscle, an important feature of the force-deformation velocity relationship F_v is that, at large lengthening velocities, the active muscle force becomes virtually independent of the velocity, i.e., a damping effect related to F_v yields zero.

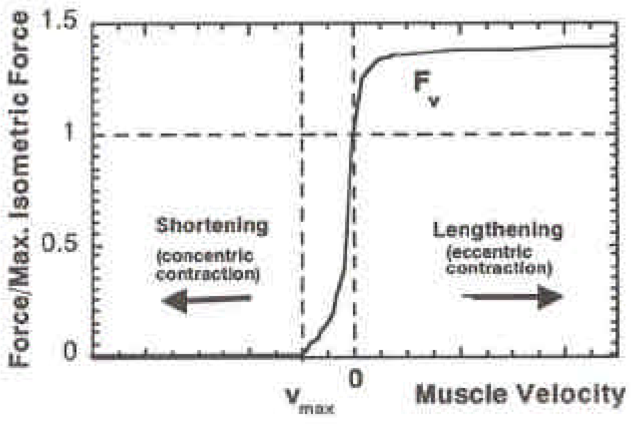


Fig. 3.8 Schematic representation of the active muscle force-deformation velocity relationship F_v. the muscle force is normalized to the maximum force obtained during isometric contraction.

The force-time relationship is determined for an isometrically contracting muscle. This relationship is typically normalized to the maximum force developed by the muscle during such a contraction (i.e., the maximum F_L value) and referred to as the active muscle state (Winters and Stark, 1985, 1987, 1988). When a segment of the human body is subjected to an unexpected external stimulus, two phases can be distinguished in a typical history of the active force generated by a muscle attached to this segment (Figure 3.9). In the first phase, changes of the active muscle force are caused only by changes in the muscle length and deformation velocity. Thus, during this phase the active muscle force-time relationship (i.e., the active muscle state) remains at its initial level. In the second phase, the force generated by the muscle increases with time, up to the maximum force.
The two phases in the active muscle force-time relationship can be explained as follows. The generation of the muscle force is controlled by the central nervous system CNS. This system senses phenomena that take place in the environment by receptors which are specialized nerve cells (Schmidt, 1978). Therefore, when an unexpected stimulus occurs, the following chain of the phenomena takes place. First, the signal from the receptor is transferred to the CNS. Then, the information transferred by this signal is processed in the CNS. Finally, the control signal from the CNS is sent to the muscle. The time delay in neuromuscular response associated with this chain of phenomena is called reflex time T_{ref} or neural delay (Tennyson et al., 1977) (Figure 3.9). The reflex time depends on the type of receptor which is involved in the perception of the phenomena in the environment, e.g., visual, vestibular or kinesthetic receptors. It is often assumed that, in a car collision, muscles are activated by the stretch reflex (Foust et al., 1973; Tennyson et al., 1977). This type of reflex involves muscle spindles which are stretch receptors located in skeletal muscles. The neural delay associated with this reflex is determined mainly by the time of conduction of the control signal between the muscle and the

spinal cord (Schmidt,1978). This implies that a short delay can be expected for cervical muscles since they are located close to the spinal cord.
When the control signal from the CNS reaches the muscles, biochemical processes responsible for muscle force generation are started, and the active muscle force begins to gradually increase with time (Figure 3.9). The dynamics of this increase is often described using a set of first order ordinary differential equation (Winters and Stark, 1985; Winters and Stark, 1987; Pandy et al., 1990). This implies that, when the time exceeds T_{ref}, the active force-time relationship can be represented by an exponential curve (Figure 3.9). The exact shape of this curve is determined by muscle mass and fibre composition. When the muscle mass and amount of slow-twitch fibres in a muscle increase, the slope of force-time curve decreases, i.e., the time necessary for a muscle to reach its maximum force increases (Winters and Stark, 1985; 1987).

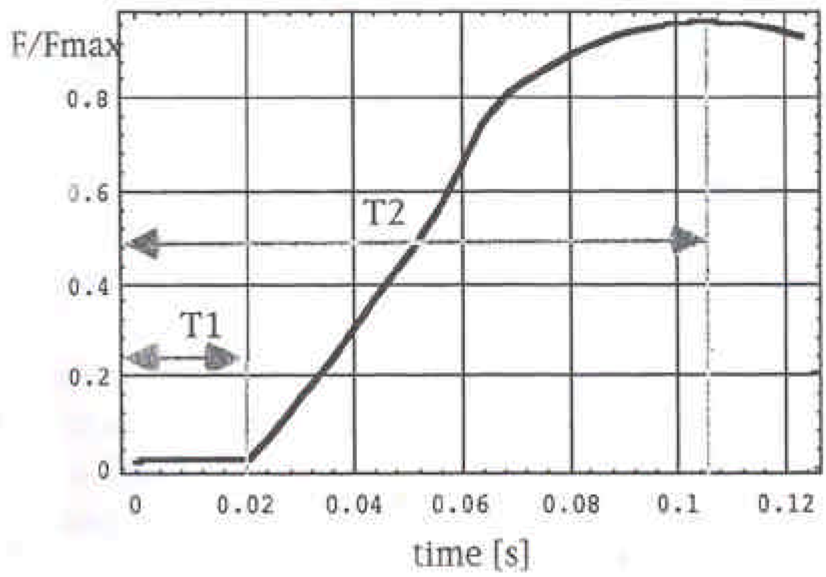


Fig. 3.9 Schematic representation of the relation between muscular force and time. T_1 is a reflex time, and T_2 is the time necessary to generate maximum muscular force.

According to this scheme, the delay between the onset of an external signal and the time that is necessary to generate maximum muscular force can be estimated as follows:
- The distance between the motoneuron and spindle (displacement sensor in muscle) of extremity muscles is about 1 m. The speed of transmission of the control signals in the nervous system is about 100 m/s (Deutsh, 1993). Hence, the time necessary to send a signal from a muscle to the spinal cord and back again is about 20 ms.
- The contraction is preceded by the propagation of a control signal (action potential) along the muscle fibre. The velocity of this propagation is about 5m/s. For large muscles with a length of around 0,4 m the time necessary to transmit a signal from the centre to the two ends is about 40 ms.
- Accordingly, the time necessary for a muscle to perform a reaction is 20 ms + 40 ms = 60 ms.
- When a contraction is triggered a certain amount of time is necessary to generate maximum force. The response of a muscle o a single stimulating signal, that is isometric twitch, can be considered as a rough simplification of contraction dynamics. In isometric twitch, muscular force reaches a maximum in around 20-200 ms (Bahler et al., 1968; Dudel, 1978; Jennet, 1989).

Therefore, total time necessary for a muscle to develop maximum force can be estimated as 80-260 ms. The description just presented for the delay between the onset of an external signal and time needed to generate maximum muscular force is based on a very simple analysis and should be regarded only as a very rough estimation. In response to danger, as in impacts, the excitability of the nervous system increases. Thus, the time necessary to perform a muscular reaction can be significantly different from that estimated in the present study.

3.3 STUDIES FOR DETERMINING REFLEX TIMES OF CERVICAL MUSCLES

In the analysis of influences of muscular forces on the kinematics of the human body segments under impact conditions, it is necessary to assess the delay time between the onset of an eternal load (for instance acceleration) and the response performed by muscles. This assessment can significantly influence the results of an investigation. Underestimation of the delay time leads to overestimation the role of muscles, and vice versa.

Surface electrodes are usually used to pick up myoelectric activity in the investigation of muscular responses and determination of reflex time of the cervical muscles when volunteers are subjected to external loads simulating those which occur in a car accident.

In the experimental literature, a reflex time is typically defined as the time lag between the stimulus start and onset of increased muscular activity represented by an increase in the EMG signal amplitude (Foust et al., 1973). In the impact biomechanics studies, such a stimulus can be generated by exposing volunteers to acceleration pulses produced in three different ways: 1) Using controlled car-crashes; 2) by means of sled devices; or 3) using acceleration pulses applied directly to the volunteers' head. The first two methods represent conditions of an actual car collision more accurately than the third one. However, application of some kind of sudden acceleration directly to the head-neck complex can simplify an experimental procedure as it requires no sled apparatus.

For instance, in the study by Foust et al., (1973), the sudden acceleration was generated by dropping a weight attached by a cord to the volunteer's head. Using surface EMG, they determined that, under this type of transient load, reflex time of the cervical muscles are in the range of 56-92 ms for the flexors and 54-87 ms for extensors. They found that these times are dependent on gender and age. As examples of the studies conducted to determine reflex time of the cervical muscles under conditions simulating those of an actual car collision, the experiments by Szabo and Welcher (1996) and Ono et al. (1997) are discussed here. Szabo and Welcher (1996) exposed their volunteers to acceleration pulses of up to 15g generated in rear-end car-to-car impacts at a speed of 16 km/h. From EMG signals recorded by means of surface electrodes, they found that the start of muscular activity occurred at approximately the same time for both cervical and extensor. Reflex time of the cervical muscles reported by them varied between 90 and 140 ms in relation to the start of the car acceleration, and between 30 and 100 ms in relation to the start of the head acceleration. In the experiments by Ono et al., (1997) the acceleration pulse was generated by means of the sled apparatus. The average impact speed was 8 km/h, and the average acceleration magnitude was 4 g. Reflex time obtained by Ono et al. (1997) were shorter than those reported by Szabo and Welcher (1996); they were in a range of 76-93 ms in relation to the sled acceleration onset.

When determining reflex times of cervical muscles under sudden acceleration it is important to answer the question of how these times may be affected the volunteers' training and their expectation of the impact. The studies reported in the literature do not give a clear answer to this question. For instance, Matsushita et al. (1997) conducted experiments under two distinct conditions of the volunteers' awareness of the impact: 1) the volunteers were kept aware of the specific time of the impact (expected impact), and 2) the volunteers were kept unaware of the impact time (unexpected impacts). They found no significant differences in the sternoleidomastoid reflex times between these two conditions: 78 and 86 ms for the expected and unexpected impacts, respectively. However, they reported that, for the volunteers who kept aware of the impact, the duration of muscular activity was longer than that for the volunteers who were kept unaware. On the other hand, Pope et al. (1998) observed that, for trained volunteers, the duration of muscular activity was shorter in the expected impact. However, they found no such difference in the activity duration when the volunteers were not trained. Thus, it can be concluded that the data reported in the literature on reflex times of the cervical muscles under unexpected external load exhibit a large variation (Table 3.1), which may be related to factors of two kinds. The first one is that the reflex times can be affected by volunteers' gender and age (Foust et al., 1973). The second kind of factors that may exert an influence on the neuro-muscular responses under unexpected external load is differences in volunteers' training and their awareness of the impact (Pope et al., 1998). The effect of awareness, however, has not been completely understood yet.

It should also be noted that an important limitation of the reported studies on reflex times of cervical muscles under unexpected external load is that they lack an exact description of the method of EMG signal processing used to determine these time. To determine a reflex time, one must detect the onset of increased muscular activity represented by increase in the EMG signal amplitude. However, in none of the studies discussed in the present section, has a precise description of a method applied to detect such an increase been given. Without such a description direct comparison of the reflex times obtained in different studies may not always be appropriate.

Reflex time		Muscles	Volunteers	Experimental procedure	Reference
From the impact onset	From the head acceleration onset				
-	59-67 ms	Cervical flexors	Females, age 18-24	Load applied the head	Foust et al. (1973)
-	72-78 ms	Cervical flexors	Females, age 62-74		
-	65-74 ms	Cervical flexors	Males, age 18-24		
-	79-92 ms	Cervical flexors	Males, age 62-74		
-	54-59 ms	Cervical extensors	Females, age 62-74		
-	54-65 ms	Cervical extensors	Males, age 18-24		
90-125 ms	40-100 ms	Cervical flexors	Group of males and females, age 22-24	Rear- end car-to-car impact at speed of 16 km/h	Szabo and Welcher (1996)
95-140 ms	65-100 ms	Cervical extensors			
76-93 ms	-	Cervical flexors	Males, age 22-34	Rear-end impact using a sled device at the speed of 8 km/h	Ono et al. (1997)

Table 3.1 Summary of the literature data on the reflex times of cervical muscles

The reflex time and the time to the peak of muscular force are derived from electromyography (EMG):

Time to the peak of muscular force = time to first peak of EMG + mechanical delay.

The mechanical delay is the delay time between the peak of EMG signal and the maximum force (Tennyson and King, 1976; Tennyson et al., 1977). The time needed for a muscle to generate maximum force can also be assessed by measurement of the relation between the acceleration of the human body segments and time. When muscular forces resist external load, the deceleration of a given segment of the body and the muscular forces reach their maximum at the same time (Foust et al., 1973). From the data available in the literature, it can be concluded that the reflex time ranges from 26 to 92 ms for the unexpected impact. The muscular force reaches a maximum in 115-251 ms after the onset of acceleration, see Figure 3.10. The delay times, which vary between muscles, also depend on impact severity. Delay times are lower for extensors than for flexors and decrease with increasing load.

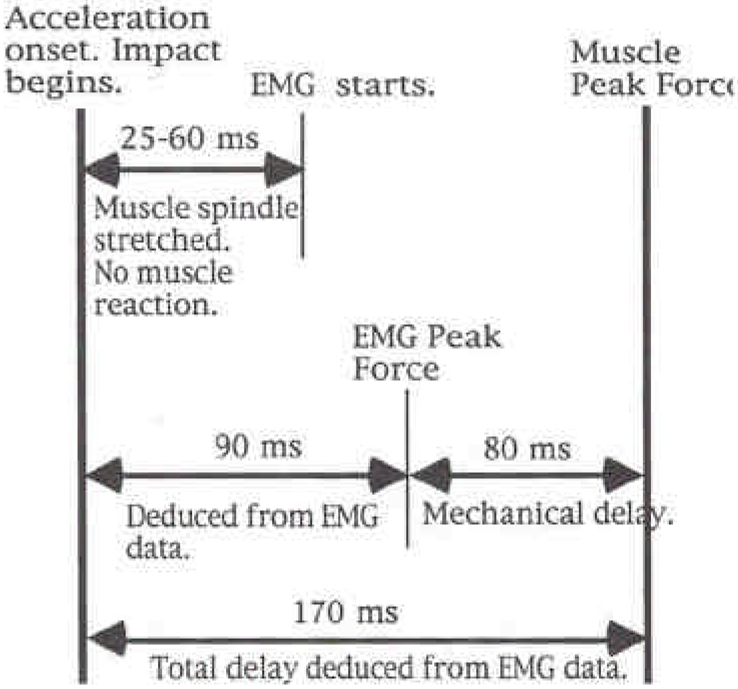


Figure 3.10 Muscle response estimated from the experimental data of Tennyson and King (1976)

In general, the delay time between the onset of an external load, for instance an acceleration pulse, and observable muscle reaction is not less than 25 ms. The first peak of EMG derived force occurs not sooner than after 45 ms. Therefore, the minimum time necessary for a muscle to generate a significant force can be estimated at 70-115 ms level. This leads to the conclusion that when the mechanical loads reach very high values in 40-50 ms, muscles cannot perform any significant reaction.

In a simplified analysis of the role of muscles in impacts, there are only two extreme cases that can be considered. The first is when muscles are not activated and develop only passive force. This case corresponds to the situation of impact that occurs so suddenly that the nervous system is not able to perform any action. This causes the role of muscles to be underestimated. The second extreme case is that of muscles in tetanic contraction when impact starts. This is equivalent to the situation when the central nervous system receives some signals from the environment before the impact. Thus, the role of muscle is overestimated. For an anticipated impact as may occur in a frontal (visual reflex), reflex time will be smaller and can even be negative if the reflex starts before the moment of impact.

3.4 STUDIES FOR DETERMINING THE STATIC STRENGTH OF CERVICAL MUSCLE

The most common method of estimation of the equivalent static strength of cervical muscles is that of determination of the maximum force applied to the head that can be resisted by these muscles in posterior and anterior directions (Mertz and Patrick, 1967,1971; Foust et al., 1973; Mayoux-Benhamou, 1989). Knowledge about such a force makes it possible to determinate a torque developed by cervical muscle around occipital condyles. The values of this torque reported in the literature are in a range of 10-32 Nm and 7-14 Nm for extensors and flexors, respectively (Mertz and Patrick, 1967, 1971; Foust et al., 1973; Mayoux-Benhamou et al., 1989). These are considerable values in comparison with the torque around the occipital condyles measured on volunteers and PMHS exposed to accelerations of up to 15 g simulating car collisions (Table 3.2).

Peak torque [Nm]	Peak sled acceleration [g]	Impact direction	Subject	Reference	Comments
10,0	2,0	Frontal	Volunteer	Mertz and Patrick (1971)	Relaxed muscles
16,5	3,9	Frontal	Volunteer	Mertz and Patrick (1971)	Relaxed muscles
30,5	6,8	Frontal	Volunteer	Mertz and Patrick (1971)	Relaxed muscles
88,1	9,6	Frontal	Volunteer	Mertz and Patrick (1971)	Tensed muscles
45,0	15,0	Frontal	Volunteer	Wismans et al. (1987)	No data on muscles tension
19,0	6,0	Rear	Volunteer	Mertz and Patrick (1971)	-
22,0	6,0	Rear	PMHS	Mertz and Patrick (1971)	-
42,0	10,0	Rear	PMHS	Mertz and Patrick (1971)	

Table 3.2 Maximum values of occipital condyles torque determined in sled tests conducted on volunteers and PMHS. Based on Mertz and Patrick (1967, 1971) and Wismans et al. (1987).

Furthermore, when the head-neck complex is subjected to such accelerations, the cervical muscles may elongate at high rates. Thus, according to F_v relationship, forces in the cervical muscles may be around 40-50% higher than those generated under static conditions, which yields torques of around 48 and 22,5 Nm for cervical extensors and flexors, respectively. Comparison of these torques values with the experimental data in Table 3.2 obtained under transient loads suggests that activated cervical muscles might exert a non-negligible effect by constraining the head motion in frontal and rear-end car collisions at acceleration of up to 15 and 6 g, respectively.

Knowledge of the maximum external forces applied to the head that can be resisted by cervical muscles does not make it possible to obtain the exact values of maximum forces generated by the individual cervical muscles. The reason is that the moment arms of these muscles are very difficult to determine in vivo. Even such advanced methods as a computer tomography seem to be deficient for determining these moment arms (Mayoux-Benhamou te al ., 1989). Therefore, a typical method of estimation of maximum static (i.e., isometric) forces F_{max} generated by the fully activated muscles is based on knowledge of muscle physiological cross-sectional area PCSA (e.g., Yamaguchi et al., 1990):

$$F_{\max} = s_{\max}.PCSA$$

Where s_{max} is the maximum isometric muscles stress. PCSA can be determined either by dissection of PMHS (e.g., studies summarized by Yamaguchi et al., 1990) or by computer tomography scanning (e.g., Mayoux-Benhamou te al ., 1989). The literature data on PCSA exhibit a large variation, which is related to anthropometric differences between PHMS (table 3.3). Similarly, the values of s_{max} reported in the literature are in a wide range of 0,25 (Herzog, 1994) to 1,0 MPa (Ikai and Fukunaga, 1970). One possible reason for a large variation of s_{max} can be that PCSA describes an entire muscle cross-section. However, F_{max} might be expected to be proportional to the cross-sectional area of the contractile tissue only. This implies that the presence of fat in muscles can be a source of error when estimating their $s_{max.}$

Thus, the current review of the literature data on the static torque developed by cervical muscles around occipital condyles suggests that, when these muscles are fully activated, they might exert non- negligible effects on the head-neck complex responses in frontal collisions at acceleration of up to 15 g. However, one can expect high variability of these effects between individuals.

The current review of the literature data on the muscle physiological cross-sectional area (PCSA) of the cervical muscles, only the stronger and more superficially located were selected because they have been considered the prime movers. While the deep cervical muscles are not included because they have been thought of as stabilizers. These muscles, which run from one vertebrae to another, do not directly influence the motion of the head, but have an indirect influence as they act to stiffen the spine.

Muscle	PCSA[cm^2]	Reference
Extensors		
Splenius capitis	1,22-3,73	Yamaguchi et al (1990)
Splenius cervicis	0,70-3,73	Yamaguchi et al (1990)
Semispinalis cervicis	2,00	Yamaguchi et al (1990)
Semispinalis capitis	2,38	Yamaguchi et al (1990)
Longissimus capitis	0,5-1,22	Yamaguchi et al (1990)
Longissimus cervicis	0,60	Yamaguchi et al (1990)
Levator scapulae	17,75	Yamaguchi et al (1990)
Multiftidus cervicis	5,80	Yamaguchi et al (1990)
Trapezius	6,45	Yamaguchi et al (1990)
	2,48	Mayoux-Benhamou et al (1998)
Sternocleidomastoid	1,60-2,08	Yamaguchi et al (1990)
Flexors		
Longus colli	0,75-0,82	Yamaguchi et al (1990)
Longus capities	0,75	Yamaguchi et al (1990)
Hyoid	0,93	Yamaguchi et al (1990)
Scalenus anterior	0,9	De Jager (1996)
Scalenus medius	1,0	De Jager (1996)
Scalenus posterior	0,9	De Jager (1996)

Table 3.3 Physiological cross-sectional areas of cervical muscles obtained from the literature

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3.5 EXPERIMENTAL STUDIES OF THE EFFECTS OF CERVICAL MUSCLE ON THE HEAD-NECK COMPLEX RESPONSES UNDER TRANSIENT LOADS

The experiments using volunteers performed by Severy et. al (1995) were done in order to determinate the kinematics of the human head-neck complex in rear-end car-to-car collisions at low speeds. Their results can be used to draw qualitative conclusions on the muscle effects on this kinematics. Severy et al. (1995) found that, in some of their experiments, the peak acceleration of the head decreased despite the increase in the impact velocity. Based on the analysis of high-speed photographs they concluded that such a decrease occurred only in the experiments in which a pronounced elevation of the upper extremity and a restraining activity of the neck muscles were observed. Their conclusion can be treated as support for hypothesis that muscles can restrain the motion of the head-neck complex in rear-end impact at low speeds.

Another study in which the muscle activity was only qualitatively estimated is that of Mertz and Patrick (1967, 1971), who simply asked the volunteer participating in their experiments either to relax or to tense his muscles. They compared the responses of the head-neck complex under these two distinct conditions of the activity of cervical muscles. Such a comparison suggested appreciable differences in the head kinematics and the maximum torque about the occipital condyles related to muscular activity in rear-end impacts at acceleration of up to 6.6 g.

Qualitative information about the muscles effects on the kinematics of the head-neck under an acceleration of a car collision can also be obtained by comparison of the kinematics of volunteers and post-mortem human subjects (PMHS). The basis for such a comparison is that the movements of the volunteers are likely to be restrained by muscle tension. Comparison of responses of the head-neck complex of PMHS and volunteers under transient loads has been made by Wismans et al. (1987). They have found that, in frontal impact with peak acceleration of 15 g, the peak value of the head flexion angle as greater in tests using PMHS than in tests using volunteers. However, Thunnisen et al. (1995) have indicated that this finding was affected by the differences in the method of measurement of the angular displacements between the tests using PMHS and volunteers in the study by Wismans et al. (1987). Their re-analysis of the results by Wismans et al. (1987) has shown that differences between PMHS and volunteer responses can exist but that they are less evident than those reported by Wismans et al. (1987).

In several studies reported in the literature, investigations of the effects of muscles on the head-neck complex responses in a car collision have been based on analysis of EMG signals of cervical muscles. For instance, in the experiments by Ono et al. (1997), volunteers were asked either to relax or tense their muscles before an impact started, and the actual state of activity of their cervical muscles was determined by means of surface EMG. Ono et al. (1997) have found that the pre-impact tension of the cervical muscles can reduce the peak value of the head extension angle by 30-40% in a car rear-end impact at a speed of 6 km/h and acceleration of around 2.5 g.

Thus, it can be concluded that the experimental studies reported in the literature have indicated that cervical muscles may affect the kinematics of the head-neck complex in rear-end impacts at low speeds (i.e., up to 16 km/h) when the volunteers anticipate an impact by tensing their muscles (Mertz and Patrick, 1967 and 1971; Ono et al., 1997). It can also be suggested that muscles may exert some effects on the head-neck complex kinematics in the frontal impacts at a high speed (i.e., around 50 km/h). This suggestion is based on the experimental results of Wismans et al. (1987) and Thunnissen et al. (1995), who reported differences in the peak values of the head angular displacement between the tests using volunteers and PMHS.

Muscle tension was more accurately controlled in the study by Verriest et al . (1972), who investigated the response of the head-neck complex of baboons on frontal impacts. Subjects were anesthetised to keep muscles in relaxed state, and electric stimulation as used to activate muscles. The results by Verriest et al. (1972) indicated that muscle tension can reduce maximum angular acceleration of the head by about 40% in frontal impact with acceleration of 20 g.

Bosio and Bowman (1986) observed phenomena which indirectly confirm that muscles can significantly influence the flexion-extension angle of the head in a car crash. They identified two modes of the head extension about the occipital condyles in frontal impacts:1) extension with rebounding after reaching a given peak of extension angle, and 2) extension without rebounding after the first peak. The extension without rebounding can be related to significant activity of the neck extensors

However, the possibility of application of the data from the literature in quantitative evaluation of the muscle effects on the head-neck complex responses in a car collision is greatly limited, as only qualitative information about muscular activity during impact tests have been given in the majority of the studies reported in the literature, e.g., Mertz and Patrick (1967, 1971) and Wismans et al (1987). It should also be noted that even the EMG signal analysis, which is a method typically used to obtain information about muscle activity impact tests, can be treated only as an indirect indicator of muscle force. The reason is that the precise, quantitative nature of the EMG-force relation of dynamically contracting muscles has not been determined yet (Herzog et al., 1994). Furthermore, direct recording of forces from individual muscles requires invasive measurement methods. Therefore, direct measurement of muscle force in volunteer under load of a crash is strongly limited by ethical and technical reasons.

For these reasons mathematical modelling are used such as alternative solution for detailed analysis of the muscle effects on the responses of the human body under transient loads.

3.6 BIOMECHANICAL FUNCTION OF A MUSCLE

The inactivated muscle has physical properties that are similar to those of other noncontractile soft tissues. The mechanical output of an active muscle is dependent upon the external load and the muscle length. The passive muscle resists, and the active muscle produces force that seems to be related to the cross-sectional area of the muscle. A representation of the active muscle function by mathematical model was proposed in 1939 by Hill. A modified Hill’s model that also included the passive behaviour of the muscle, according to Fung, is diagrammed in Figure 3.11.

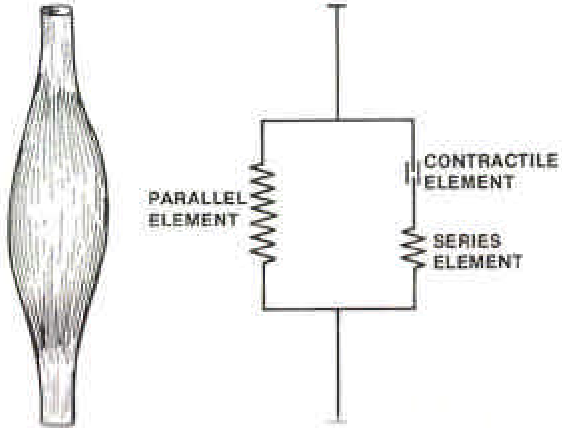


Fig. 3.11 Functional model of muscle. Physical properties of a muscle are quite different when it is passive or in active state. Both these aspects of muscle behaviour may be represented in a quantifiable manner by the three-element mathematical model shown on the right. The model consists of a parallel element, representing the passive elastic behaviour of the muscle, and a series element which, together with a contractile element, represents the active elastic behaviour of the muscle. The parallel and the series elements have constant stiffness values for a given muscle, while the contractile element is variable depending upon the activity of the muscle. Such models for individual muscles may be incorporated into the mathematical models of the spine to represent the total active behaviour of the entire spine.

The model consists of three elements, two spring-like elastic elements (parallel and series), and one contractile element under the control of a neuromuscular signal. The passive behaviour of the muscle is completely represented by the parallel elements, since the contractile element is inactive, and therefore no force is transmitted by way of the series element. When a muscle is voluntarily contracted, it may either remain in a fixed position with no change of muscle length (isometric contraction), or it may contract and shorten (isotonic contraction) to provide work against external load. In both situations, the series element shares the load together with the parallel element. This effectively increases the muscle stiffness. It should be emphasized that the mathematical model presented in Figure 3.11 is not a physical representation of a muscle, but it is a simple and precise way to describe the actual mechanical behaviour of the muscle.

4. REVIEW OF THE MATHEMATICAL MODELLING OF THE MUSCLE AND THEIR EFFECTS ON THE HEAD –NECK COMPLEX

Mathematical modelling plays an important role in classical and impact biomechanics for three main reasons. The first is that reliable and detailed models make it possible to calculate deformation, stress, and force in the human body tissue. The second reason for application of mathematical models in impact biomechanics is that such models validated against experiments on PMHS can predict the human body responses under conditions resulting in high risk of injuries and fatalities. Furthermore, mathematical models are an efficient tool for conducting parameter studies, which is the third reason for their application in impact biomechanics. Such model yield fully repeatable results and their behaviour is completely determined by the set of their input parameters. Thus, it can be concluded that validated mathematical model might be regarded as a method for supplementing and better understanding experimental results obtained using volunteers and PMHS.

4.1 MODELLING OF MUSCLE INFLUENCE ON THE KINEMATICS OF THE HEAD-NECK COMPLEX IN A FRONTAL CAR IMPACTS

For this analysis it has been used the simplified model of the head complex that consisted of the rigid head and neck segments connected by a revolute joint. In this model, the system of cervical muscles was represented by means of three groups of flexors and three groups of extensors (figure 4.1 and table 4.1)

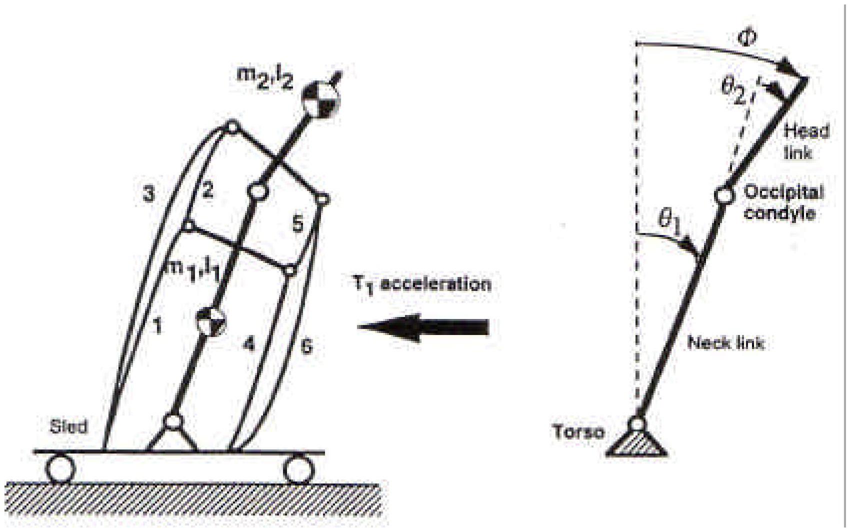


Fig. 4.1 The current model of the head-neck complex with muscle elements (curved lines). On the left model with muscle elements (curved lines), see also table 4.2. on the right definition of flexion angles. Positive angle is flexion; negative one is extension. m_1 and m_2 are the lumped masses of the neck and head, respectively; I_1 and I_2 are the mass moments of inertia of the neck and head, respectively. T_1 is the first thoracic vertebrae.

Muscle element	Muscles
1	Splenius cervicis, semispinalis cervicis, longissimus cervici, levator scapulae
2	Longissimus capitis, semispinalis capitis
3	Trapezius
4	Scalenus
5	Lungus capities, longissimus capities
6	Sternocleidomastoideus

Table 4.1 Muscle elements from the model shown in Figure 4.1.

Maximum isometric torques/ forces of muscles were calculated from the literature data. Maximum static torque generated by flexors was assumed to be 50% lower than the maximum static torque generated by extensors. Maximum force developed by given muscle was calculated as a value proportional to the physiological cross-section area of the muscle. Physiological cross-section areas of the neck muscles were estimated from the data summarized by Yamaguchi et al. (1990). For more details about the parameter used in this analysis to see [24].
The muscles were simulated by means of Hill- type model, see Figure 4.2.
The structure consists of contractile element and series elements, they describe the active elastic behaviour of the muscle. The parallel element describes the passive elastic behaviour of the muscle.

Fig. 4.2 Structure of muscle model. CE is contractile element, SE is the series contractile element, PE is the parallel elastic element.

Influence of muscles at different impact severity and for different reflex time. Three impact pulse were to analyse influences of muscles at different impact severity, see Table 4.2. Pulses T12 and T30 correspond to mean value of horizontal T_1 acceleration-time histories in the NBDL frontal impact tests at peak sled acceleration 6 g and 15 g, respectively. Pulse T70 is based on an upper bound of T_1 horizontal acceleration in the NBDL frontal impact test at 15 g. This pulse was used to simulate a very severe impact.

Peak sled acceleration	Peak T_1 acceleration	Symbol in text	Description	Reference
6 g	12 g	T12	Mean values from volunteer tests	Allen and Bowman (1986)
15 g	30 g	T30	Mean value from the most severe volunteer tests	Wismans et al. (1994)
15 g	70 g	T70	Envelope of volunteer tests	Wismans et al. (1987)

Table 4.2 Characteristic of impact pulses used in the current study.

Eight values of the T_{reflex} ranging from 0 to 0,2 s (0,00 s; 0,02 s; 0,04 s; 0,06 s; 0,08 s; 0,10 s; 0,12 s; 0,20 s) were used at each impact pulse. T_{reflex} = 0 corresponds to the situation when the nervous system reacts immediately after an impact onset. A reflex time lower than 0,06 s can be regarded to be a "fast" reaction of the nervous system. Since only initial 0,2 s of the impact were analysed for the impact pulses T30 and T70, T_{reflex} corresponds to the situation when no muscle action is performed during the impact. At the pulse T12 the head flexion angle reached maximum after 0,2 s. for this reason 0,3 s of an impact were analysed for this pulse. In this case T_{reflex}= 0,3 s was used to model the situation when no muscle action is performed . The choice of values for the reflex time is based on results of the previous literature review (Wittek and Kajzer, 1995).

Influence of muscles on response of the head-neck model greatly decreased when the impact severity increased. The maximum angle of the head was reduced by up to 65% at the impact pulse T12, and only by about 20% at the impact pulse T70, see Figure 4.3 and 4.4. At impact pulses T30 and T70 relative influence of muscles dropped to a value lower than 15% when T_{reflex} was greater than 0,12 s, see Figure 4.4.

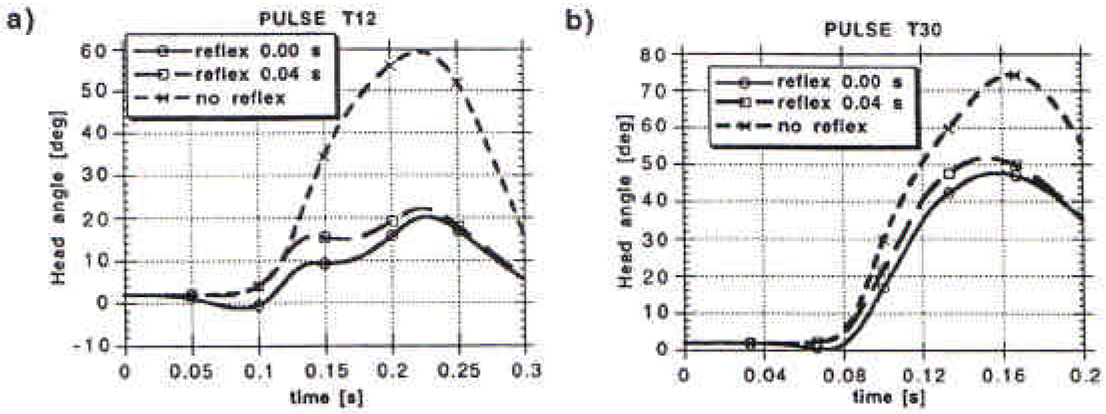


Fig. 4.3 Head flexion angle-time histories at the different impact severity and for different reflex time. a) load pulse T12. b) load pulse T30.

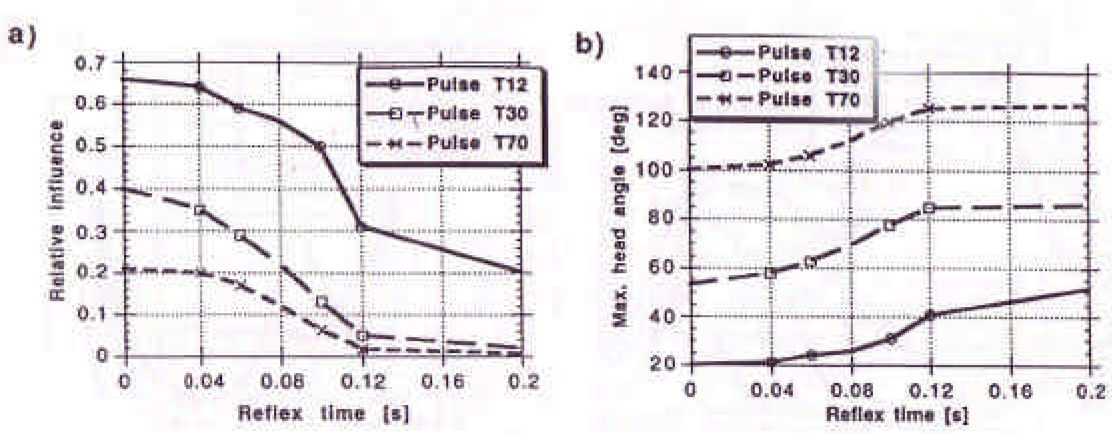


Fig. 4.4 Influence of muscles on the head flexion angles as a function of the reflex time at different impact severity. a) relative changes of the maximum angle of head flexion. b) the maximum angle of head flexion.

Influence of reflex time and initial muscle tension. The influence of the reflex time on the response of the model of the head-neck complex was analysed in more detail at the impact pulse T30. The time histories of the head flexion angle, and the head angular acceleration calculated for different reflex times were compared (for more details to see [24]).

When the reflex time was lower than 0,04 s, muscles decreased the maximum angle of head flexion by about 40% at the impact pulse T30, see Figure 4.5a. However, the maximum angular acceleration of the head was reduced by muscle tension only by about 10% see Figure 4.5b.

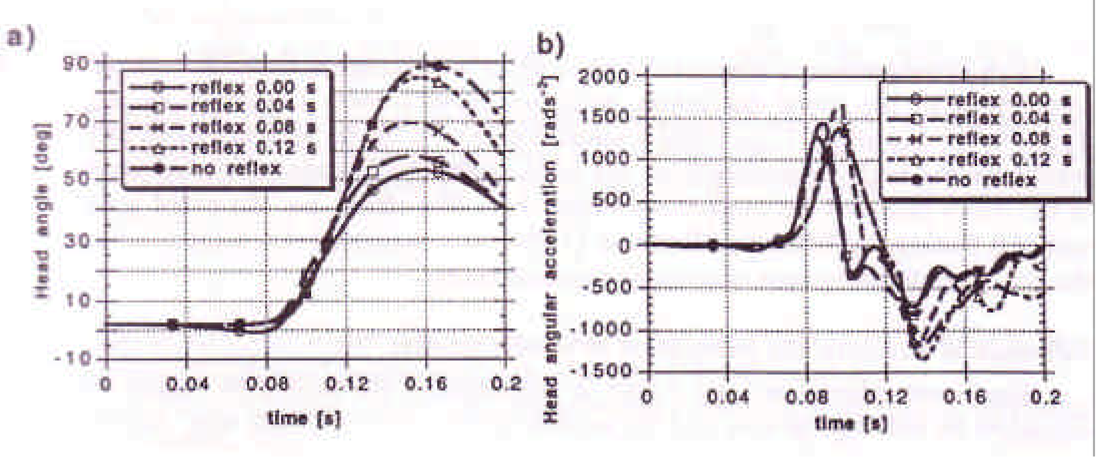


Fig. 4.5 Results for impact pulse for different values of the reflex time. a) angle of the head flexion b) angular acceleration of the head.

For the reflex time greater than 0,10 s the maximum force extensors was calculated to be lower than 0,8 F_{max}, and relative influence of muscles was lower than 20%. Furthermore, the force extensors, quickly dropped after the first peak, see Figure 4.6. In consequence, muscles could not significantly affect kinematics of the head-neck complex for high values of the reflex time.

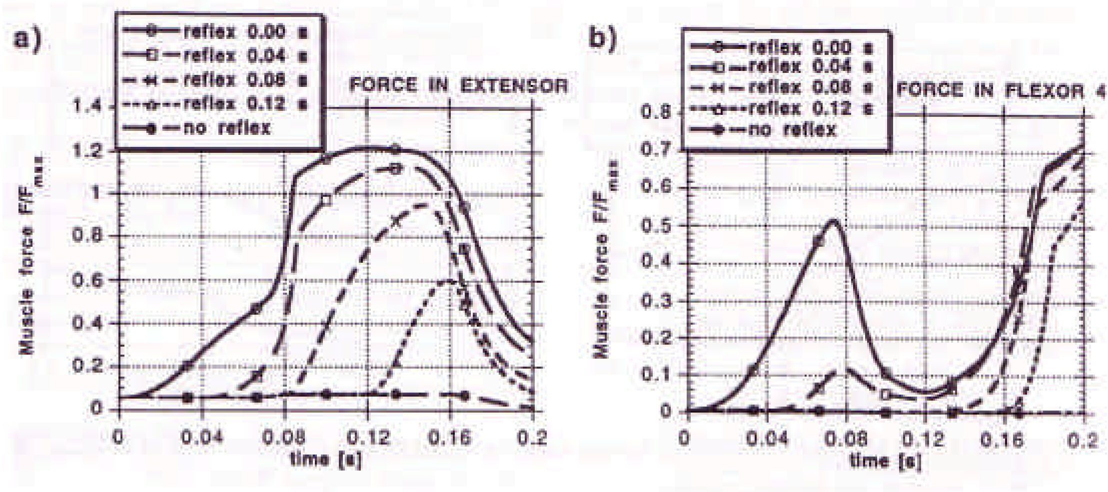


Fig. 4.6 Muscle-force time histories for different reflex time. a) muscle 1 (extensor), b) muscle 4 (flexor). Force is normalised to F_{max}.

Influence of the static neck strength. In the current study, the static neck strength was defined as the maximum static torques generated by the muscles about the neck-torso joint and about the occipital condyles. Based on the literature review, the following data on the maximum static torque were selected for investigation of the influence of the static neck strength:

1) 30 Nm for neck extensors and 15 Nm for head extensors;
2) 65 Nm for neck extensors and 30 Nm for head extensors;
3) 100 NM for neck extensors and 50 Nm for head extensors.

Maximum torque of neck extensors equalled 30 Nm can be regarded as a low value which corresponds to elderly persons. Whereas a value 100 Nm corresponds to young, well-trained subjects. Data from point 2 are an average of the literature data.

Effect of muscle force on the maximum angle of head flexion was almost linearly dependent on the static neck strength, see Figure 4.7. Maximum angles of the head flexion were 36°,53° and 73° for the maximum static torques of the neck extensors 100 Nm, 65 Nm, and 30 Nm, respectively.

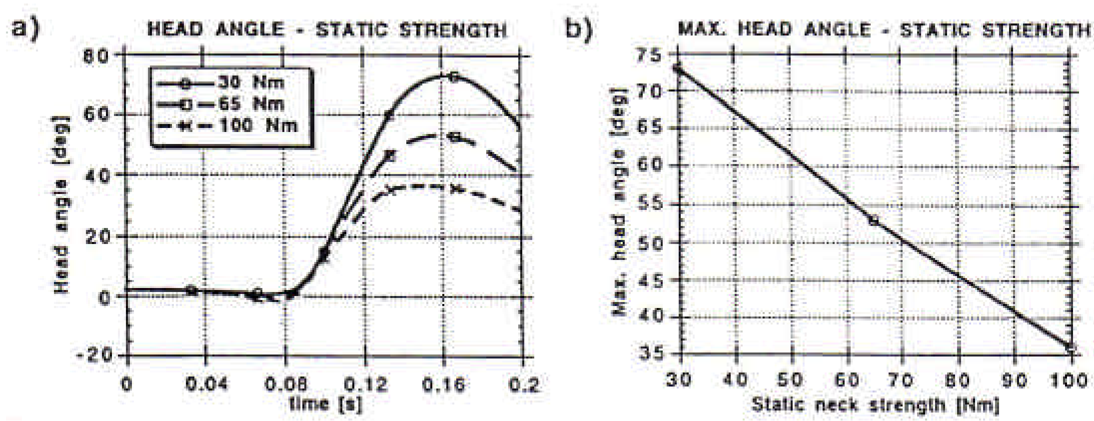


Fig. 4.7 Influence of the static neck strength pulse T30. In this figure, the term static neck strength is referred to the maximum static torque of extensors. a) head flexion angle-time histories for different values of the static neck strength. b) maximum angle of head flexion as a function of the static neck strength.

Conclusion. The muscles can significantly affect kinematics of the head-neck complex in 15 g frontal impact when the following conditions are satisfied simultaneously:

1) peak of T_1 horizontal acceleration is less than 70 g;
2) reflex time is lower than 60 (80) ms.

When these conditions were satisfied, the maximum angle of the head flexion was decreased by muscle tension by up to 40%. When the peak of T_1 acceleration was about 70 g, muscles were not able to significantly influence kinematics of the head-neck complex, even for very low values of the reflex time. It can be concluded that the muscle effect is negligible in frontal impacts which are more severe than 15 g.

The results of the current study indicate that kinematics of the head-neck complex strongly depends on the assumed initial tension in muscles. Therefore pre-impact activity of the neck muscles should be taken into account in the analysis of the muscle effect on the kinematics of the head-neck complex in a car crash.

The static neck strength was identified as a biomechanical parameter which has the strongest effect on the calculated maximum angle of head flexion was almost linearly dependent on the static neck strength.

In the summary, the present analysis indicates that muscle effect may be negligible at impact pulses which are more severe than 15 g and for reflex times greater than 100 (120) ms.

4.2 FINITE ELEMENT MODELLING OF THE MUSCLE EFFECTS ON KINEMATICS RESPONSES OF THE HEAD-NECK COMPLEX IN FRONTAL IMPACT AT HIGH SPEED

The aims of this study were as follows. Firstly, to validate the model of the neck with implemented muscle elements by comparing its kinematics under an acceleration simulating a frontal car collision at high speed with literature data on responses of volunteers and post-mortem human subjects (PMHS) exposed to such an acceleration. Secondly, to utilize this validated model to perform a parameter study of the muscle effects on the head-neck complex kinematics in such a collision. This parameter study was focused on understanding how dependent are the calculated muscle effects on two factors: 1) reflex time of the neck muscles; 2) distribution of the maximum isometric forces within the system of these muscles.

The impact load was represented by a horizontal acceleration of the first thoracic vertebra T_1 according to the results by Wismans. The acceleration utilized here correspond to the average results of sled tests in which volunteers were subjected to a frontal acceleration pulse with a magnitude of around 15 g. The impact velocity in these tests was around 60 km/h.

Neck muscles were modelled using the Hill-type multi-bar muscle element. In this muscle element, a skeletal muscle was represented as a string of series-connected bars, see Figure 4.8. The behaviour of these bars was determined using the Hill-type muscle model. The force in each bar F_{musc} was calculated using the following formulae:

$F_{mus} = F_{Act} + F_{PE}$ and

$F_{Act} = F_{\max} N_a(t) F_L(L) F_V(V),$

where F_{Act} is the active muscle force, F_{PE} is the passive muscle force, L is the muscle length, V is the velocity of muscle deformation, N_a is the muscle active state, F_L is the muscle force-length characteristic (at V=cost and $N_a(t)=1$), and F_{max} is the muscle maximum isometric force, for more details to see [24].

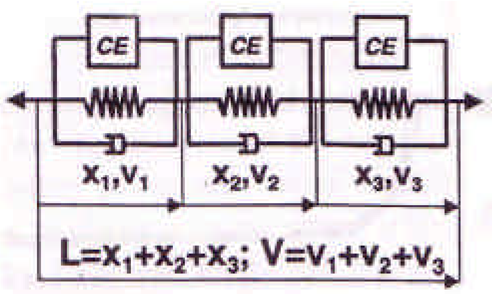


Fig. 4.8 Scheme of muscle force calculation in the current Hill-type multi-bar muscle element. CE is the contractile element. $X_{i=1,2,3}$ and $v_{i=1,2,3}$ are elongation and rate of elongation of bar *i* in the multi-bar muscle element, respectively. L and V are total elongation and rate of elongation of the entire multi-bar muscle element, respectively.

Attachment point of the neck muscles and the muscle geometry were estimated based on the literature. All muscles which have been considered in the literature to play an important role in the control of movements and position of the head-neck complex were taken into account: six flexors (longus colli, longus capitis, scalenus anterior, scalenus medius, scalenus posterior, and hyoid) and ten extensors (sternocleidomastoid, splenius capitis, splenius cervicis, semispinalis capitis, semispinalis cervicis, longissimus capitis, longissimus cervicis, levator scapulae, multifindus, and trapezius).
Almost all of them were grouped according to different lines of muscle force action, which resulted in 86 multi-bar muscle elements (Figure 4.9). In the current analysis, the sum of the neck muscle cross-sectional areas was 19,6 cm^2 for extensors and 8,0 cm^2 for flexors. We selected the peak muscle stress s_{max} of 1 MPa, which yielded 49 Nm of the maximum static torque of cervical extensors around the head-C_1 joint.

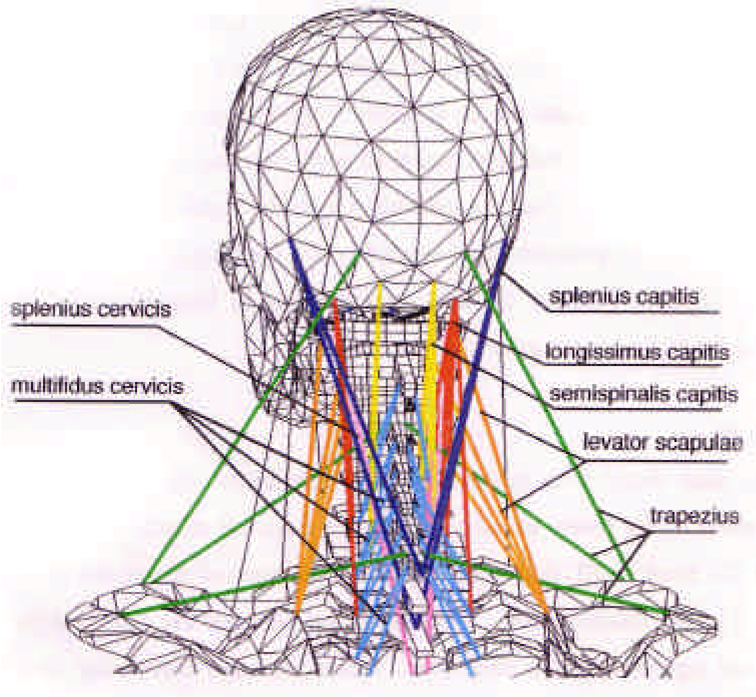


Fig. 4.9 Posterior view of the modified model of the neck with muscle elements. Neck extensors are plotted using solid lines and named.

Influences of distribution of maximum isometric force on the calculated muscle effects. In order to estimate the possible influences of the distribution of the maximum isometric forces F_{max} on the calculated muscle effects, the active force F_{Act} of the neck extensors was concentrated in the splenius capitis and splenius cervicis muscles, it was assumed that the sum F_{max} of these two muscles equals the sum of F_{max} of all the neck extensors. The reason for selecting the splenius capitis and splenius cervicis was they are long extensors of the neck with large cross-sectional areas. When the active force of the neck extensors was concentrated in these two muscles, the maximum static moments developed by them at the initial position of the neck around the head-C_1 and T_1-C_7 joints were higher than those calculated using anatomical distribution of $F_{max.}$ The calculated muscle effects were stronger by around four times than those obtained for the anatomically distributed F_{max} (Figure 4.10).

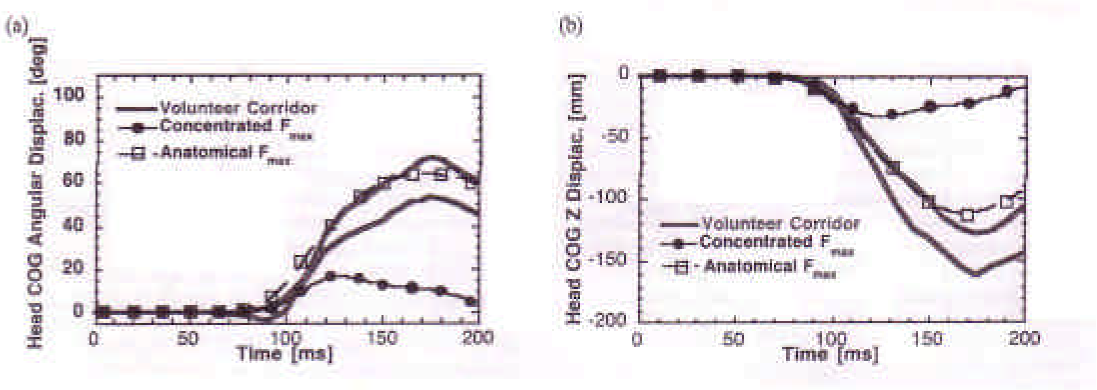


Fig. 4.10 Comparison of the results calculated under the assumption that the static strength of the neck extensors is concentrated in splenius capitis and splenius cervicis (concentrated F_{max}) with those obtained using anatomically determined isometric forces of the neck extensors (anatomical F_{max}). a) Head COG angular displacement and b) Z displacement of the head COG.

When the model with F_{max} of extensors concentrated in this way was computed under the assumption that reflex time is 50 ms, the peak value of the calculated angular and Z displacement of the head gravity center were clearly below the lower boundaries of the experimental corridors obtained by Thunnisen using volunteers. On the other hand, when the maximum isometric forces of the neck extensors were determined using the muscle physiological cross-sectional areas, the head angular displacement agreed well with the experimental results of Thunnissen (Figure 4.10).
Concentric the total strength of the neck extensors in the splenius cervicis and splenius capitis implied that large muscular forces were exerted directly on the head and upper part of the neck, which appreciably increased muscle moments around the head-C_1 and T_1-C_7 joints. Thus, the current results suggest that an appropriate estimation of the maximum isometric force (i.e. PCSA) of large cervical muscles is an important factor in predicting the muscle effects on the kinematics responses of the head-neck complex in a high-speed car collision.

Muscle effects for different reflex times. We investigated the muscle effects on head-neck complex kinematics in a frontal impact for four reflex times (25 ms, 50 ms, 80 ms and 100 ms). We assumed that all the neck extensors start to react at the same time. In the current study, a reflex time was defined as the time lag between the impact acceleration onset and the time when muscle active state started to increase. According to experimental results of Szabo and Welcher, the neck muscles begin to actively react to around 90-140 ms after the start of lumbar spine acceleration.
The muscle effects on the head-neck model responses greatly increased when reflex time decreased. This is because for low reflex times, T_{reflex}, the muscle elements started to generate

forces sooner, and the peak forces were higher than those calculated for long T_{reflex} (Figure 4.11). In consequence, for T_{reflex} of 25 ms, time history of Z displacement of the head gravity center of the current head-neck model was clearly below the corridor of the results of experiments using volunteers reported by Thunnissen.

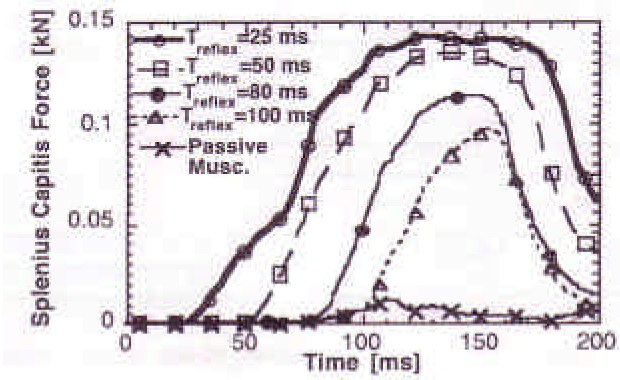


Fig. 4.11 Splenius capitis, force calculated for different reflex times T_{reflex} and for the passive muscles.

The peak values of the calculated angular and Z displacements of the head gravity center were about two times lower than those obtained under the assumption that the muscles were inactivated (Figure 4.12). When the reflex time was assumed to be 50 ms, the calculated displacements decreased by around 35-40% in comparison to those calculated for inactivated muscles. Thus, for T_{reflex}=50 ms, both the angular and translational displacements of the head gravity center, and its angular acceleration were very close to the boundaries of the experimental results obtained using volunteers reported in literature (Figure 4.12). When the reflex time was increased to 100 ms, the muscles affected the peak values of the calculated angular and Z displacements of the head gravity center by only around 10%, this is an almost negligible effect. In consequence, for both inactivated muscle elements and the elements activated at 100 ms, the peak values of the calculated angular and Z displacements of the head gravity center were beyond their corridors obtained in the tests using volunteers (Figure 4.11).

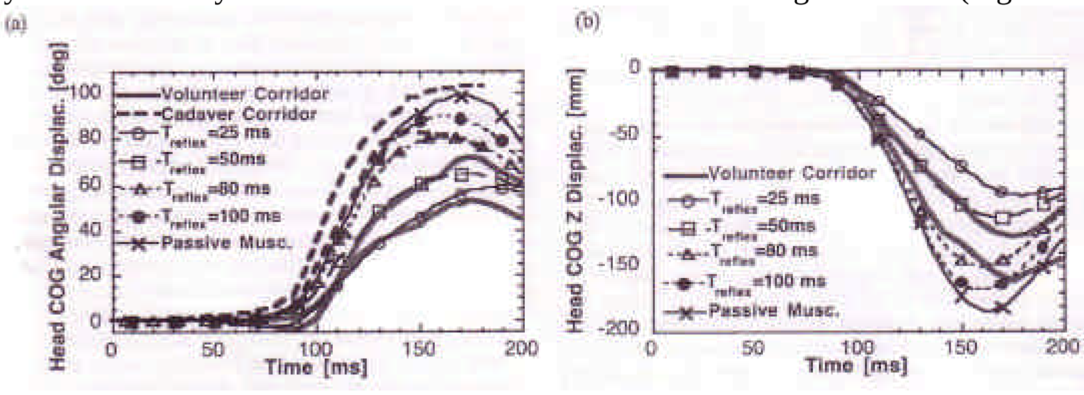


Fig. 4.11 Comparison of responses of the current neck model with muscle bars with the experimentally obtained corridors reported in the literature. a) head COG angular displacement and b) Z displacement of the head COG.

Conclusion. The current analysis of the muscle effects on the head-neck complex responses for various reflex time does not allow one to clearly determine the threshold for neck muscle reflex time beyond which these effects become negligible. However, it suggests that the neck muscles are likely to exert signification effects on the kinematics of the head-neck complex subjected to a frontal impact with a peak acceleration of 15 g when they are activated at around 25-50 ms after the start of such an impact. On the contrary, when the activation starts after 100 ms or later, muscle effects can be disregarded. This model suggests that muscles may significantly

decrease these displacements when they begin to generate an active force at around 25-50 ms after the start of the first thoracic vertebrae acceleration. Moreover, it can be presumed that muscles may decrease the risk of injuries to the neck as they are likely also to reduce the angular acceleration of the head gravity center.
The results discussed here, they showed that the concentration of such forces in strong and long muscles attached to the skull, such as the splenius cervicis, and splenius capitis, can lead to an overestimation of the muscle effects.

4.3 FINITE ELEMENT MODELLING OF KINEMATIC AND DYNAMIC RESPONSES OF CERVICAL SPINE IN LOW-SPEED REAR-END COLLISIONS

The goal of the current study is to develop a mathematical model of the cervical spine that accurately represents the kinematics of the individual cervical vertebrae during a low-speed rear-end collision and the effect of muscle activity on this kinematics. The finite element model of the cervical spine with muscles, previously applied in the study (4.2) to investigate the effects exerted by muscles on the kinematics of the head-neck complex in a high-speed frontal impact, was modified to improve its bio fidelity in low-speed rear-end impacts, for more details to see [24].
The impact load was represented by the accelerations of the first thoracic vertebrae T_1 according to the results of the experiments using volunteers presented by Ono. The peak values of the sled acceleration were around 2,5 and 4 g for impact speeds of 6 and 8 km/h.
In the present model of the cervical spine, the geometry of the cervical muscles and their attachment point were estimated based on Soames (1995), Seireg and Arvikar (1989), and Yamaguchi (1990). Seven flexors and nine extensors were modelled in a way similar to that described by Wittek (1998), Figure 4.12. Almost all of the muscles were subdivided into several portions with different lines of action, which resulted in 90 multi-bar muscle elements. To simulate the curving of long flexors and extensors around the vertebrae during cervical spine bending, splenius capitis, splenius cervicis, and longus colli were modelled using three series-connected bars in each multi-bar element.

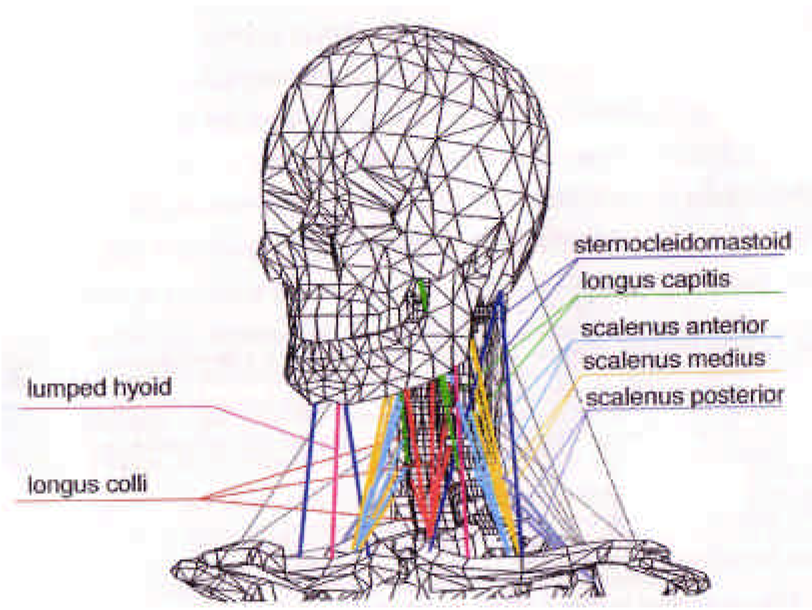


Fig. 4.12 Model of the cervical spine with muscles utilized in the current study. The cervical flexors and the sternocleidomastoid are printed in colour and named. The extensors are visualized using solid black lines.

General kinematics of the head-neck complex. In the modelling of unexpected impacts at the speed of 6 km/h, the head COG angular displacement-time history were close to the lower limit of the experimental corridor (Figure 4.13).

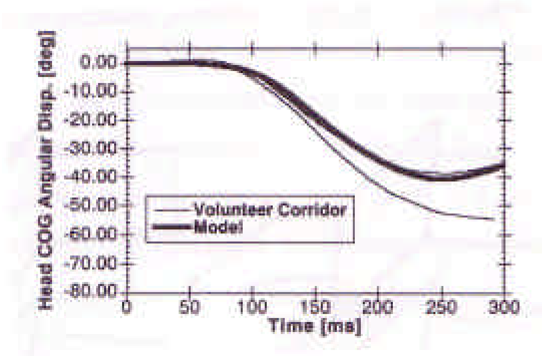


Fig. 4.13 Comparison of the calculated and experimentally obtained angular displacement of the head gravity center for unexpected impacts at speed of 6 km/h.

The magnitudes of the head COG angular and resultant accelerations agreed well with those obtained experimentally (Figure 4.14).

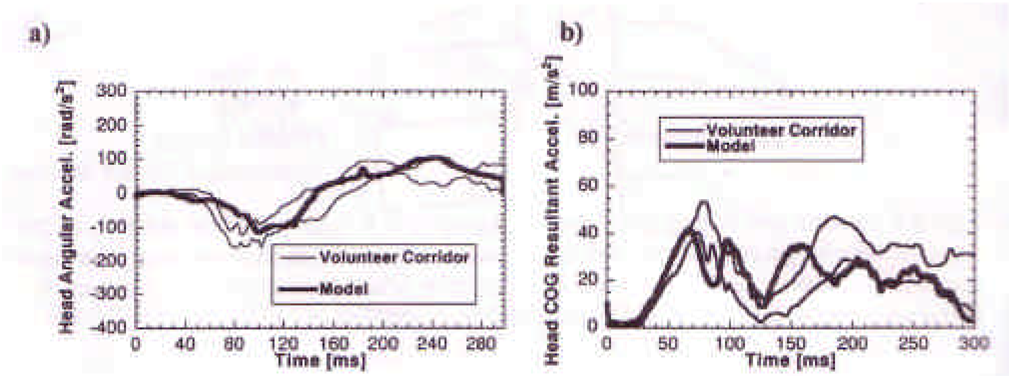


Fig. 4.14 Comparison of the results of the modelling of unexpected impacts at the speed of 6 km/h with experimental data. a) angular acceleration of the head gravity center; b) resultant acceleration of the head gravity center**.**

In the modelling of expected impacts, the calculated head COG angular displacement was in the middle of the experimental corridor (Figure 4.15). Its maximum value decreased by around 30% in comparison to that computed on the assumption that the cervical muscles were activated by stretch reflex.

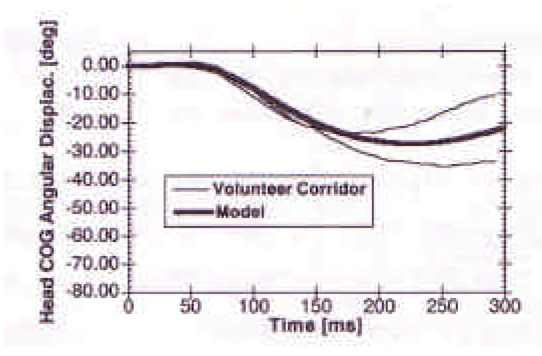


Fig. 4.15 Comparison of the calculated and experimentally obtained angular displacement of the head gravity center for an expected impact at the speed of 6 km/h

The peak values of the head COG resultant and angular accelerations were calculated to occur sooner than those obtained while simulating unexpected impacts (Figure 4.16). The maximum values of the calculated angular and resultant accelerations were lower by around 40% than those obtained in the experiments.

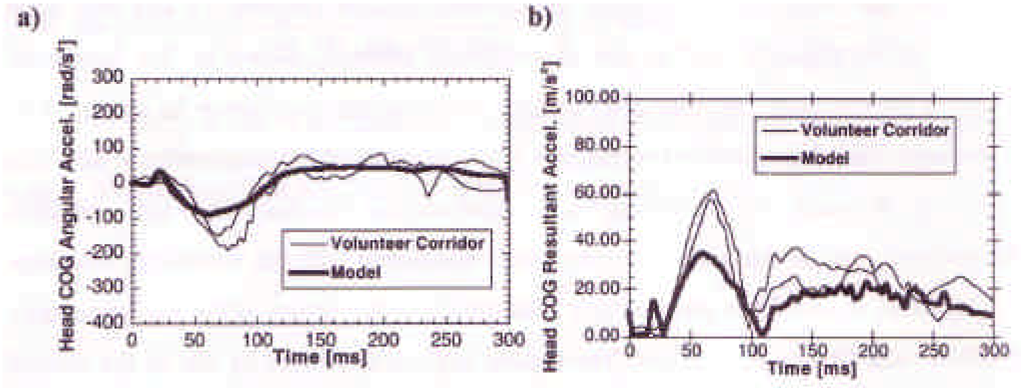


Fig. 4.16 Comparison of the results of the modelling of expected impacts at the speed of 6 km/h with the experimental data. a) angular acceleration of the head gravity center; b) resultant acceleration of the head gravity center.

Muscle effect on the general head-neck complex kinematics. Muscle tension significantly decreased the peak value of the calculated angular displacement of the head gravity center. The modelling of expected impacts yielded a head COG angular displacement of around 70% of that obtained while simulating unexpected impacts (Figure 4.17). Moreover, the peaks of the angular and resultant accelerations of the head gravity center calculated to occur 20-50 ms sooner in expected than in unexpected impacts (4. 18). One possible explanation for this shift in phase of the acceleration time histories is an increase in the stiffness of the head-neck complex caused by muscle tension in expected impacts.

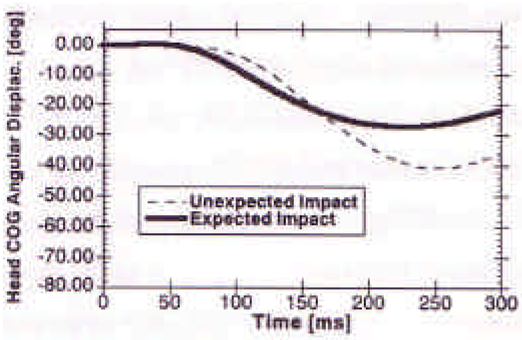


Fig. 4.17 Comparison of the angular displacements of the head gravity center calculated when modelling of expected and unexpected impacts at the speed of 6 km/h.

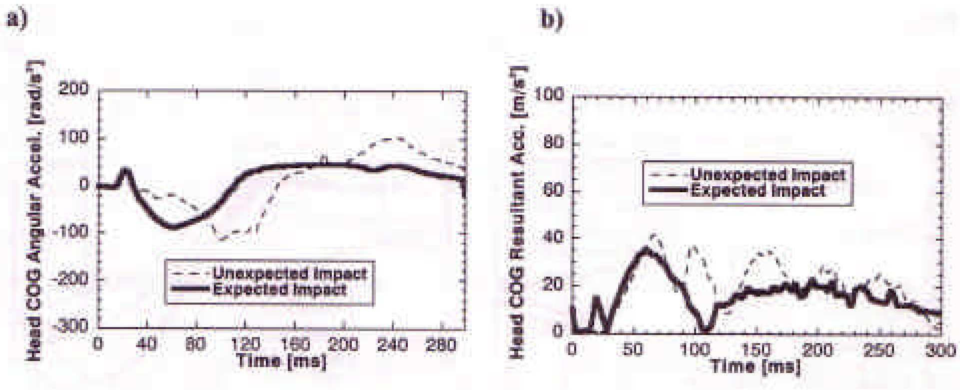


Fig. 4.18 Comparison of the angular and resultant accelerations of the head gravity center calculated when modelling of expected and unexpected impacts at the speed of 6 km/h. a) angular acceleration; b) resultant acceleration.

Effect of muscle tension on angular displacement of cervical vertebrae. A decrease in the angular displacement of the head gravity center in the expected impacts discussed in the previous section can be explained by effect exerted by muscles on the relative vertebrae angles. The modelling of expected impacts yielded relative angles between the cervical vertebrae lower than those calculated when muscles were assumed to be activated by stretch reflex (Figure 4.19). The largest relative decrease in the vertebrae angle caused by muscle action was calculated to occur at C_2-C_3 level where such angle went down by around 2° (about 40%). The relative decrease at C_5-C_6 level was only around 7%. One possible explanation for such a relative decrease in the C_2-C_3 angle might be that the major cervical flexors (e.g., longus capitis) are attached to the skull and upper cervical vertebrae.

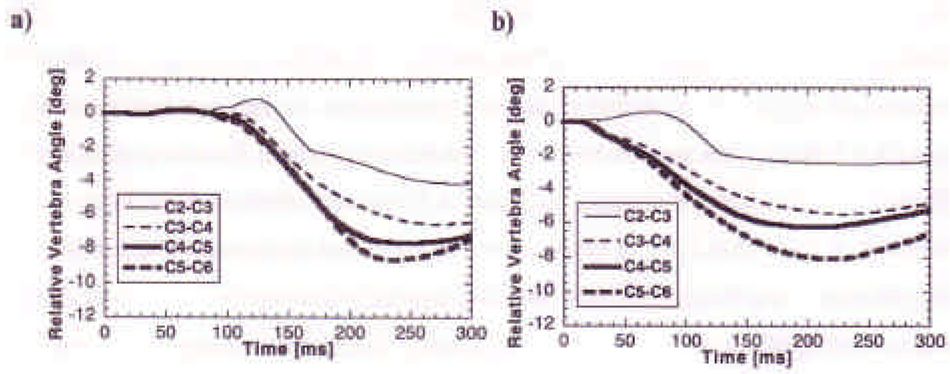


Fig. 4.19 Comparison of the relative vertebrae angles calculated when modelling of a) unexpected and b) expected impacts at the speed of km/h.

Conclusion. Validation against the general kinematics of the head-neck complex indicated that current model of the cervical spine with muscle yields angular displacement and resultant acceleration of the head gravity center close to the experimental results obtained using volunteers who kept their muscles relaxed before the start of an impact. However, for expected impacts, the calculated maximum values of the head gravity center accelerations were lower than those recorded in the tests using volunteers.
A comparison of the results calculated when modelling the expected and unexpected impacts indicated that muscles can reduce the relative vertebrae angles.

5. ODD-NECK II: MUSCLE COMPONENTS

The aim of this thesis, as mentioned earlier, is to study the human muscle system with its properties and behaviours in order to find some possible solutions to replicate these characteristics. These are for instance passive and active muscle force, the relation between force and time of contracting muscle, time to reflex, time necessary to generate maximum muscular force and other properties mentioned in the previous chapter. Besides this dummy neck has the ability to move with a motion similar to the human neck because it is able to perform flexion and extension but also lateral bending and rotational in the same time. This is because this dummy neck will be used not only in the frontal and rear-end crash test but also in lateral crash test.

The objective of this chapter is to illustrate the possible solutions for the realization of the muscle system, the new components and the modification that will be insert in the prototype. The whole ODD-neck with muscle components it is shown below (Fig. 5.1), each component is described (in the following pages) after and all detailed drawings of each piece are presented in the Appendix.

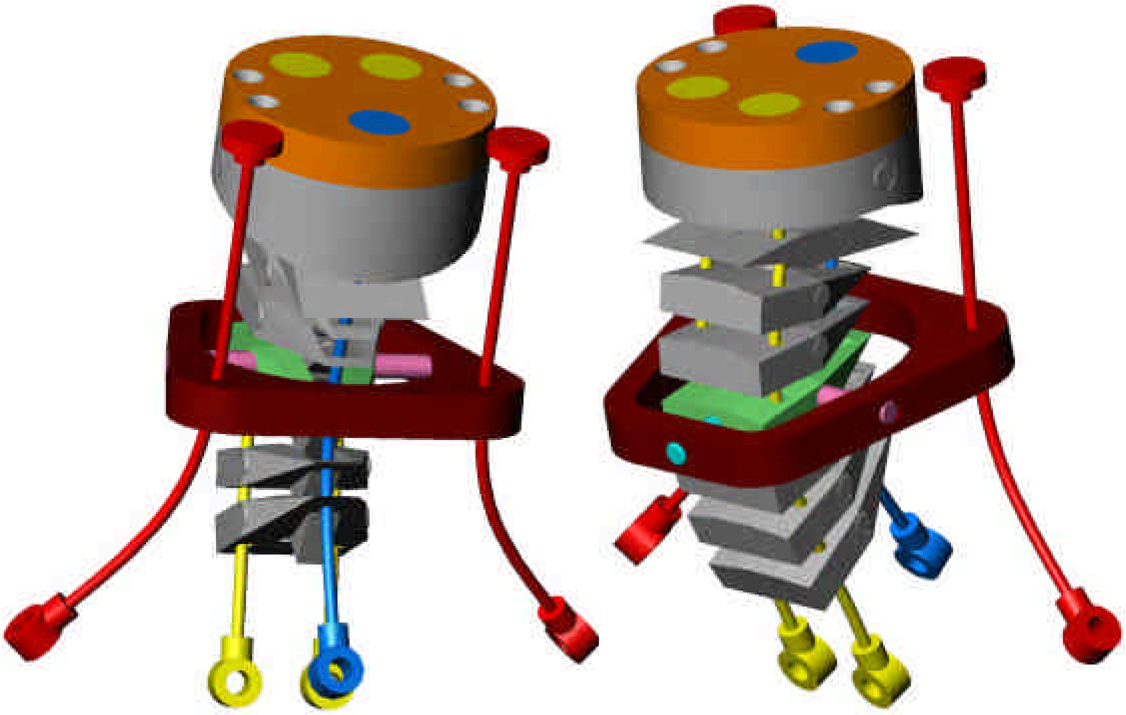


Fig. 5.1 The whole view of the 3D model of the dummy neck with news muscle components (in colour are plotted the news components, in grey the pieces already existing).

5.1 NEW COMPONENTS

The interface between the head and occiput where the cables are attached has a circular shape with three holes where the cables are fixed (Fig.5.2). This component is linked with the occiput by means of four oval machine screws (these screws haven't been realized because they are standard components).

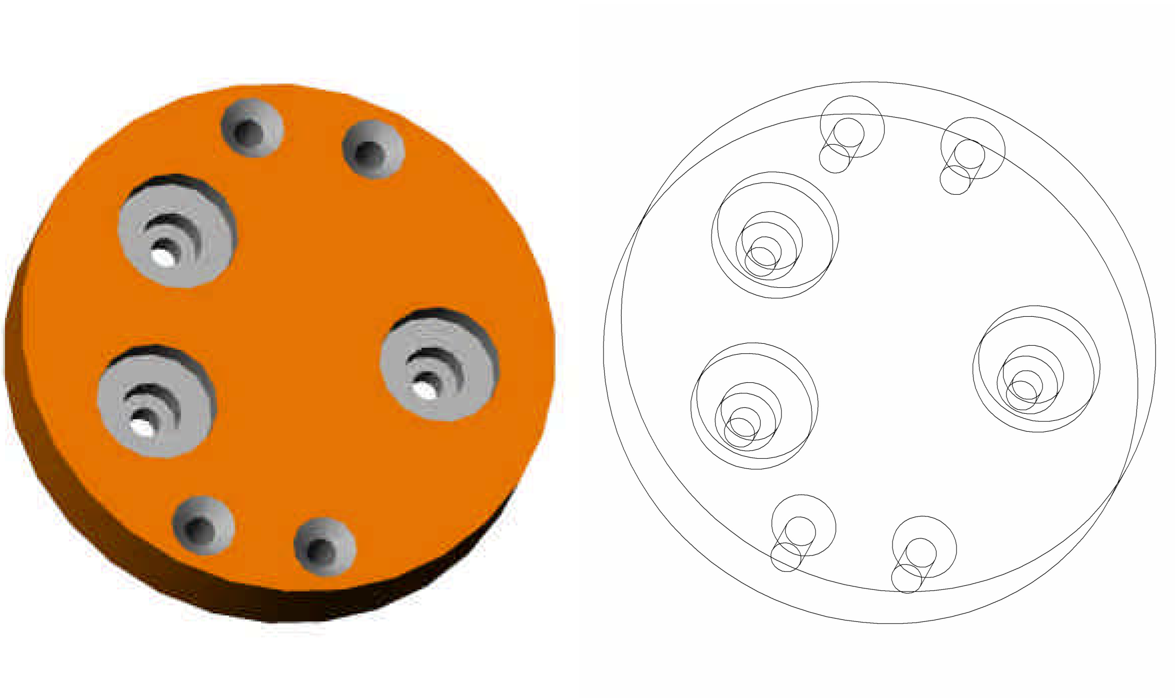


Fig. 5.2 Interface between the head and occiput, view: 3D model, wireframe.

It is possible to make this component in acetal plastic because this material has low weight but is very strong and at the same time it is easy to shape. It is possible to modify the piece in function of the kind of the interface that there is in the head.

The muscle system has been replicate by means of cables. In this prototype there is no space to replicate the whole muscle system of the human neck, but, based on literature review, only the main muscle that have a greater influence in the neck response have been reproduced. The stronger and more superficial muscles have been chosen because they have been considered the prime movers. Instead the deep muscles are not included because they have been considered as stabilizers of the head, the muscles that run from one vertebrae to another, and don't directly influence the motion of the head. The kind of the muscles that have been chosen are written in the table 3.3.

Next step to plan the cable is selecting the material to be used. Steel has been chosen, because it has a high value of the Young's modulus (the elastic behaviour of the muscle will be replicated inside the cylinder system).

The stress within the cable is described as follow:

$$s = E.e$$

Where s is the stress (N/m^2), E is the Young' modulus, for the steel is 210000 N/mm^2, e is the strain.

But the stress is:

$$s = \frac{F}{A}$$

Where F is the force (N) in this case is the maximum dynamic force in the muscle (the value has been chosen in the literature review), A is the cross section of the cable (m^2).
The strain is:

$$e = \frac{\Delta L}{L}$$

Where ? L is the change in length (m) and L is length (m).

The final equation for to find the cross area of the cable is:

$$\frac{F}{A} = E.\frac{\Delta L}{L}$$

The values choose for the different parameters are:

E= 210000 N/mm^2

F= 980 N (to see next chapter for explanation)

L= 250 mm (this is a hypothetical value, only in the final plan it is possible to know the exactly value because it depends where the cylinder will be attached)

? L= 1 mm

The equation for to find the diameter is:

$$A = \frac{p.d^2}{4}$$

The theoretic value of the diameter is 1,21 mm. A value of diameter equal to 4 mm has been chosen, otherwise making the connection with the other components is impossible. (Fig. 5.3).

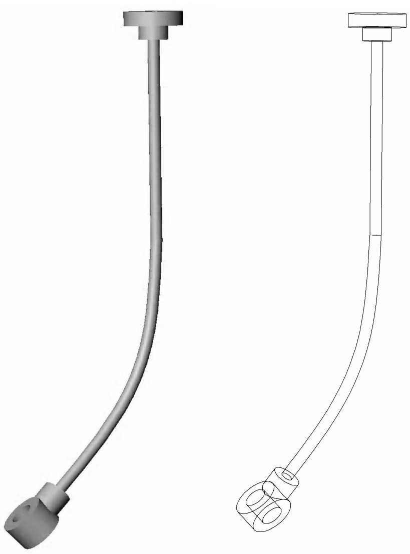


Fig. 5.3 cable: 3D model, wireframe.

The choice of putting two cables in the front and three behind at the neck is because the force in extensor muscles is higher than in flexor muscles, since the centre of gravity of the head has an anterior location. So the extensor muscles counteract gravity whereas the flexor are helped by gravity.
For reason of space, it isn't possible to put all cables within the vertebrae. It is feasible to modify the number of the external cables and also the position, because in this situation I think they guarantee a good stability in flexion-extension, but I don't know in lateral movement. Only a real test or the study with dynamic programs like MADYMO or ADAMS can give some answers (for problems of time in this thesis is not possible to study the dynamic system, but it could be done in a future work).
The cables within the neck are connected from the interface occiput-head to the cylinder located near the first rib; the external cables are connected from the head to the cylinder located between the second or third rib.
The kind of attach (connection) drawn are only to give an idea of realization but it can be modified in function of the kind of cylinder and of attach that will be choose.

In a mathematical program the muscle system is represented with a straight line from the head to the ribs, but in the human body the muscle system runs along the neck and bends before attaching to the ribs. Therefore to replicate this muscle bend, at the level of the fourth vertebrae (C_4), a piece has been inserted (Fig. 5.4). The new component has been fixed with two pins. It is possible to make this piece in acetal plastic and the two pins in stainless steel.

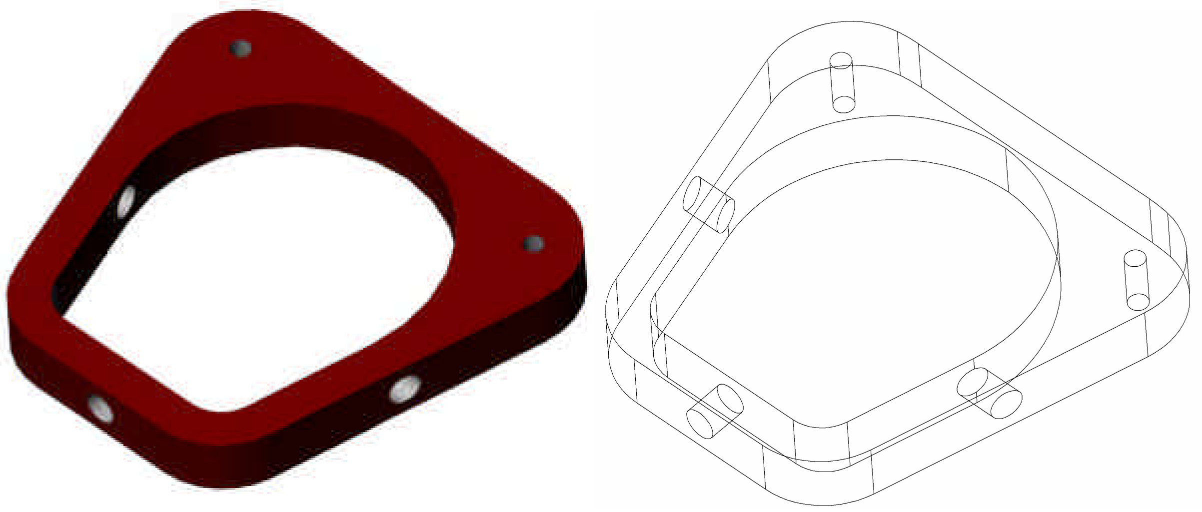


Fig. 5.4 Piece that gives the bending to the muscles: 3D model, wireframe.

5.2 MODIFICATION AT THE PIECES ALREADY EXISTING

The already existing components have been modified in order to be crossed by the cables.
In the occiput (C_0) three holes and four thread holes have been made, the latter used to fix the occiput with the head-neck interface (Fig.5.5).

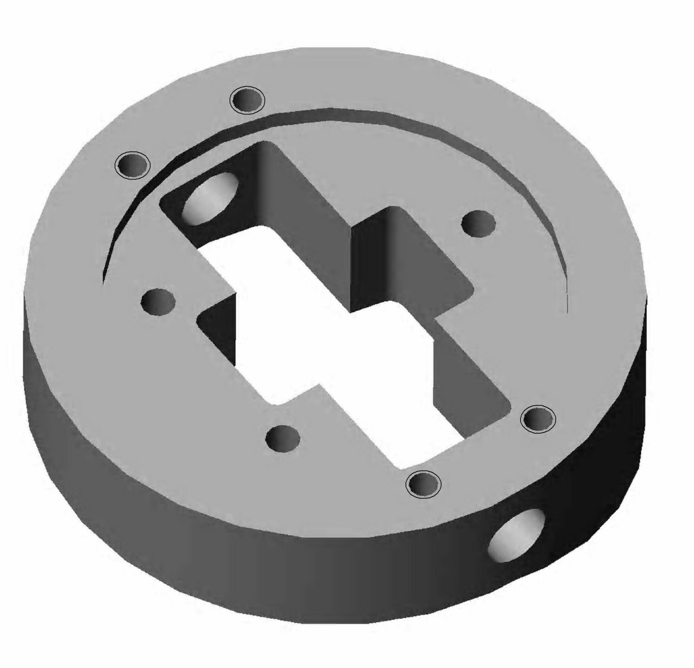


Fig. 5.5 The occiput (C_0): 3D model

In the first vertebrae (C_1) three holes have been made (Fig. 5.6).

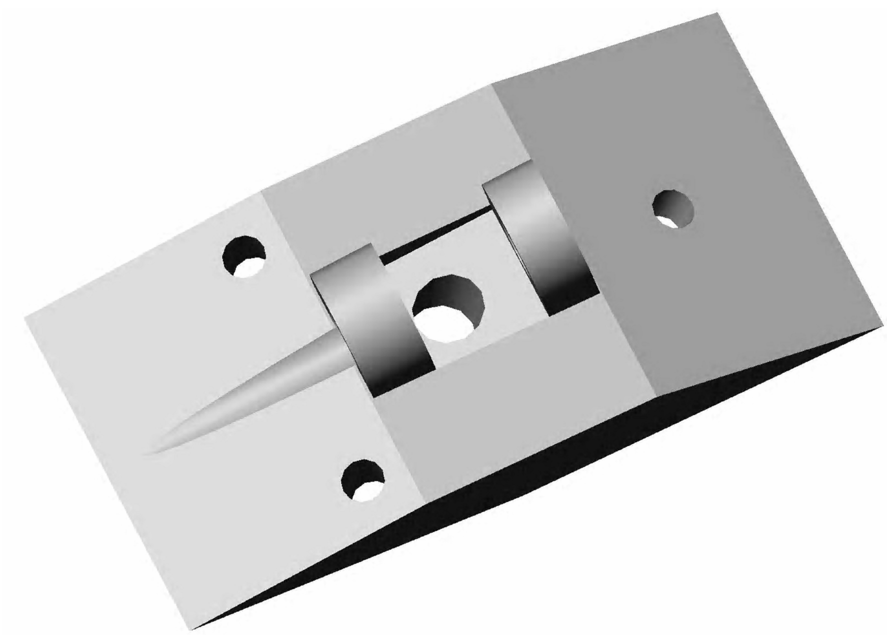


Fig. 5.6 Fist vertebrae (C_1): 3D model.

The vertebrae from C_2 to C_7 are all similar so three holes have been made in each vertebrae; only the C_4 has one hole more in the front, where the component is fixed to give the bending at the cables (Fig. 5.7).

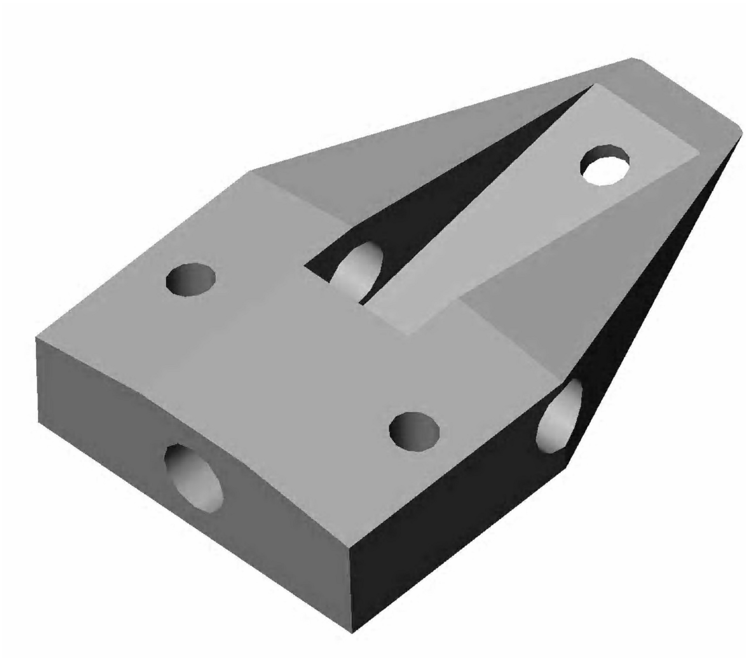


Fig. 5.7 Vertebrae C_4: 3D model.

6. ODD-NECK II: ANALYSIS OF DIFFERENT TYPES OF ACTIONAMENT

In this chapter the different possibilities to move the cables with characteristics and properties of the human muscle system have been analysed. Pneumatic, hydraulic and magnetic system have been considered.
The basic physiological property of the muscle tissue that have been considered are:

1) time to reflex, the value of this time is within a range of 26-92 ms for the unexpected impact;
2) the time of muscle contraction to reach maximum force is in the range 115-251 ms after the onset of acceleration. This means that the time necessary for a muscle to generate a adequate strength is estimated between 70-115 ms until a maximum of 150-170 ms. Therefore the displacement of the piston has to be within that range of time;
3) passive and active muscle behaviour. The muscle is only able to contract and is not able to stretch alone consequently, for instance, a movement of flexion-extension antagonist muscle are activated, as explained in the third chapter;
4) the force-deformation velocity relationship. The dynamic force in the cervical muscles may be around 40-50% higher than that generated under static condition. It is very difficult to choose an exact value of muscle strength to plan the prototype muscle system. In literature there are different values of muscle cross sections, torques around head gravity centre, and around occipital condyles. The reason is that the moment arms and the cross sections are very difficult to determine in vivo. Even if advanced methods are used such as computer tomography, there are large variations of literature data because they depend from the method utilized to analyse, the kind of people analysed, if more trained or not, the age, the weight, the height.

In the current analysis the following data have been utilized:
cross-sectional areas for extensors (PCSA) = 19,6 cm^2
cross-sectional areas for flexors (PCSA) = 8,0 cm^2
the peak muscle stress s_{max} = 1 MPa
maximum static torque of cervical extensors around the head-C_1 joint ($M_{st\ max}$)= 49 Nm
These data have been taken in the mathematical modelling previously described (see chapter 4.2). The force values obtained have been compared with others force values found in others mathematical modelling or others studies, and it has been seen that they have the same order size.
With this data the dynamic strength in each cable for extensors and flexors muscles has been evaluated. The method used is:

$$F_{st\ max}\,extensors = s_{\max}.PCSA_{extensors}$$

$$Moment\ arm\ extensors = \frac{M_{st}\max}{F_{\max}extensors} = 25\text{ mm}$$

The dynamic torque is 1.4-1.5 higher than static torque therefore, the maximum dynamic torque of cervical extensors around the head-C_1 joint ($M_{dyn\ max}$) = 73.5 Nm

$$F_{dyn\ max}\,extensors = \frac{M_{dym}\max}{moment \cdot arm \cdot extensor} = 2940\text{ N}$$

The maximum static torque cervical flexors is about the half of the maximum static torque of cervical extensors so is $M_{st\ max}$ = 24,5 Nm

$$F_{st\ max}\,flexors = s_{\max}.PCSA_{flexors}$$

$$Moment\ arm\ flexors = \frac{M_{st}\max}{F_{\max}\ flexors} = 30\text{ mm}$$

The dynamic torque is 1.4-1.5 higher than static torque therefore the maximum dynamic torque of cervical extensors around the head-C_1 joint ($M_{dyn\ max}$) = 36,75 Nm

$$F_{dyn\ max}\,flexors = \frac{M_{dyn}\max}{moment\cdot arm\cdot\ flexors} = 1200\text{ N}$$

There are three cables reproducing the extensor muscle system therefore the maximum dynamic force in each cable is 980 N
There are two cables reproducing the flexor muscle system therefore the maximum dynamic force in each cable is 600 N

5) The weight of the human chest is around 15 kg, thus the total sum of all components that will be introduced doesn't have to exceed this value, otherwise it will not have really biofidelity with the dummy and the crash test;
6) The range of change of the muscle length is 70-150 mm. This value of range has been calculated considering the results of the head gravity center angular displacement obtained in a mathematical modelling. The range depends from the severity of impact pulse, different reflex time, initial muscle tension and other parameters previously seen in chapter four.

6.1 PNEUMATIC SYSTEM

Pneumatics systems use pressurised air to make things move (for example, pistons), to hold things up (for example car tyres), or to move things (for example, air lines inside hopper feeder tubes). In the pneumatic systems there are sources of energy, methods for transporting that energy, and then methods for expending that energy. Compressors take atmospheric air and quash it, creating air at high pressure (a compressor is basically a large pump, operating at high speed: it takes air of volume x, and reduces this volume by factors y, increasing the pressure proportionately). Strong pipes then carry this high pressure air to where it is needed. This air can then be used in a number of ways. In this case a pneumatic cylinder converts air pressure into linear motion.
Pneumatic air cylinders are pneumatic linear actuators driven by pressure differential in the cylinder chambers. One side of a piston flange may be pressurized to provide force and motion with a spring providing return force after pressure is released (single action), or both sides can be alternately pressurized for bi-directional powered motion (double action).
In the pneumatic circuits, the purpose of a valve is to control the current of air to the pneumatic cylinder in such a way that the cylinder moves either inwards or outwards. Such a valve can either be operated by hand, pneumatically, or electromagnetically.

6.1.1 Pneumatic circuit

The pneumatic circuit used to activate the cables is very easy, a double action cylinder controlled by an electromagnetic valve. The cylinder has a magnetic switch, the sensor is used to detect the piston's position inside the cylinder. All these components are inside the dummy, instead the other components that create the energy are outside the dummy (for instance compressor, filter, cistern etc..) and the pressure air is carried to valve by means of the pipes (Fig. 6.1).

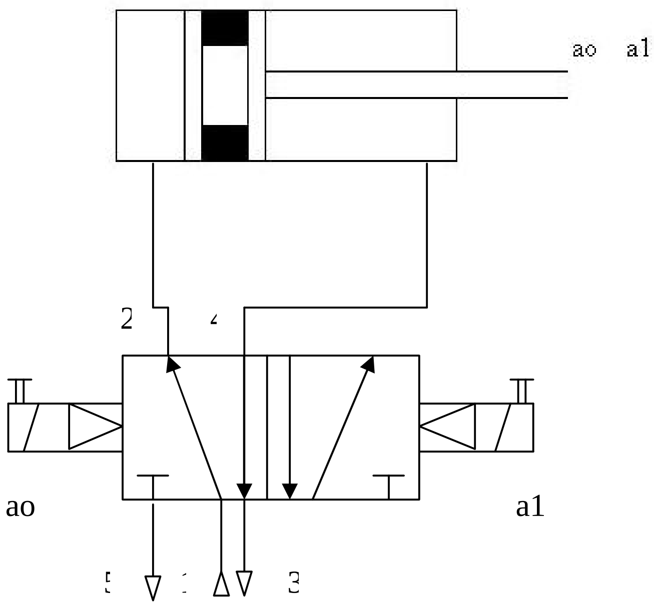


Fig. 6.1 Pneumatic circuit inside the dummy: cylinder with magnetic sensor, electromagnetic valve.

The compressor takes atmospheric air and quashes it, creating air at high pressure, then it sends the pressure air at the valve by means of the pipes. The air arrives at the valve in the point 1 (Fig. 6.1), goes through the valve, goes into the chamber of the cylinder and drives the piston outward (a_1), the air inside the other chamber goes out by means of the exhaust port, point 3. The magnetic sensor inside the cylinder detects the piston's position and sends an electric signal at the control unit (PC or PLC) that analyses and elaborates the information. After the control unit transmits an electric signal that switches the valve, the piston goes back in (a_0). The representation of this process is in the diagram below (Fig. 6.2).
The directional control valve in the pneumatic circuit should be mounted as close as possible to the cylinder. Pressure drops in the long hoses leading to the cylinder may cause the piston to move too slowly. Also, long hoses between the valve and cylinder may contain more air volume than the cylinder will accommodate, thus the lubricated air might never reach the cylinder before it is fully extended. Short pipes between the valve and cylinder will help to solve both of these problems.

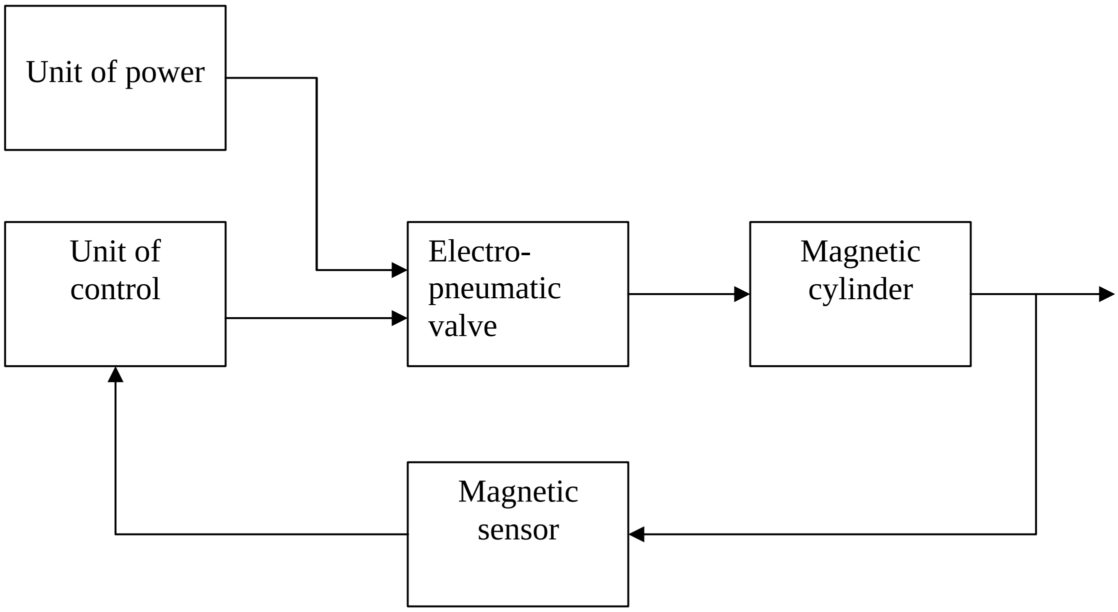


Fig. 6.2 Diagram of the process

6.1.2 Dimensioning of the pneumatic circuit

The air pressure used within the pneumatic circuit is between the range 2-10 bar, but in the exercise normal pressure used is 6 bar.
The formulae used for the dimensioning of the pneumatic cylinder are:

- Force in push

$$F = \frac{\left(\frac{p \cdot D^2}{4}\right) \cdot p}{100}$$

- Force in pull

$$F = \frac{\left(\frac{p \cdot D^2}{4} - \frac{p \cdot d^2}{4}\right) \cdot p}{100}$$

Where F is the force in kg, D is the bore in mm, p is the pressure within the pneumatic circuit in bar and d is the rod diameter in mm.
Using the force for the muscle extensors equal to 980 N (about 100 kg) and the pressure equal to 6 bar, the cylinder bore is 46 mm. It is better to choose a bore equivalent to 50 mm because, when the piston is pulled, the whole surface minus the surface of the rod have to be considered.

Using the force for the muscle flexors equal to 600 N (about 61 kg) and the pressure equal to 6 bar, the cylinder bore is 35 mm. It is better to choose a bore equivalent to 40 mm because, when the piston is pulled, the whole surface minus the surface of the rod has to be considered.

The formulae for the evaluation of the wear air is:

$$Q = 60 \cdot \frac{\left(\frac{p \cdot D^2}{4}\right) \cdot c \cdot (p+1)}{1000000 \cdot t}$$

Where Q is the carrying capacity in n_litre/min, c is the stroke in mm, p is the pressure in bar and t is the time to cover the stroke in sec.
It is important to consider an effective wear air equivalent about 40-50% higher than the theoretical value because of wear air in the pipes and in the directional control valve.
Having the value of the carrying capacity, it will be possible to select the kind of compressor used in the pneumatic circuit.

The stroke of the cylinder is like the muscle length, with a maximum stroke equal to 170 mm.
Therefore if the time necessary for a muscle to generate an adequate strength is between 100-170 ms, the speed basic profile is the trapezoidal move and the move time is divided equally into 3 parts. The first part is acceleration, the second constant velocity and the third part deceleration. The peak speed of the cylinder is 1.5 m/s and the acceleration is 18 m/s^2 almost 2g.

6.1.3 Choice of the pneumatic components

In basis at all data previously evaluated, now the best cylinder and valve suitable for the circuit must be chosen.
The main characteristic of the cylinder and valve are:
1) the cylinder has to have a magnetic sensor that detects the piston’s position inside the cylinder, the valve has to be activated electromagnetically. It is important that these two components have this kind of feature because the sum of feedback signal time, the time of analysis and elaboration data, the time for the switching of the valve and for the beginning of the piston moving have to be within the range of the time to reflex (26-92 ms). Therefore only electromagnetic components can have a time of answer in the order of the millisecond;
2) the cylinder bore for muscles extensors of 50 mm and for muscles flexors of 40 mm;
3) the cylinder stroke between 70-150 mm, this depends in which position the cylinder will be fixed (stroke is the distance between fully extended and fully retracted rod positions);
4) the peak piston speed is 1,5 m/s. In this case it is important to find a particular system for piston deceleration without crash because slowly down of the piston is very fast;

5) it is important to consider the weight of the cylinder and of the valve. They have to be made in light materials like aluminium or other special polymers;
6) it is important also to consider the maximum dimension of the valve and the cylinder because there is a limited space in the dummy.

There are a lot of companies in the world that produced valves and pneumatic cylinders. One possible solution is show below, probably there are better solutions but it is impossible to check all companies and then these kinds of products are continually evolving, therefore it is better always to check if there are news products.

Compact cylinders. The compact cylinders (series ADM of the GENERALMATIC company) meet the need to make clamps that are quite small. In fact, their main characteristic is that their overall dimensions at stroke 0 are very small compared to normal cylinders and therefore can lock or move short distances even in limited spaces. All this is obtained rationalizing component construction, considering also that the rod guide can be shorter than that of a normal cylinder. There are various versions: double and single acting, with magnetic piston push-pull rod. Double action with magnetic piston has been chosen.
They are built so that the cylinder works perfectly, even with non-lubricated air, due to the fact that the seals are self-lubricating and the body is made from anodized shaped aluminium.
On the cylinder body there are cavities where the magnetic sensors are placed and a complete range of clamps makes them easy to install under any condition.

Construction characteristics:

Body	Hard anodized aluminium
Rod	Stainless steel
Piston	Aluminium
Rod bushing	Bronze + PTFE
End plate	Anodized aluminium
Piston seal	Polyurethanes PU 90° Sh A
Rod seal	Polyurethanes PU 90° Sh A

Technical characteristics:

Fluid	Filtered and preferably lubricated air
Max pressure	10 bar
Working temperature	-30° ÷ +80°C
Weight cylinder Ø 50 mm	650 gr
Maximum dimension	50 x 67 mm and a length of 170 mm
Weight cylinder Ø 40 mm	600 gr
Maximum dimension	40 x 58 mm and a length of 170 mm

Rod guide: the front head of the cylinder, being firmly fixed to the tube, allows an excellent rod guiding and the possibility to manufacture cylinders with maximum strokes up to 400 mm (according to the cylinder bore).
All cylinders are supplied with elastic damping device of end stroke on the piston.
The weight is not exact, but can vary around that value because it depends from which kind of clevis will be used to fix the cylinder. Also the length of the cylinder depends from the stroke that will be chosen.

Electromagnetic valve. The solenoid valves (series 2000 of the PNEUMAX company) have been developed to meet requirements for electronically controlled pneumatic systems. They have been designed to be easily assembled into groups or manifolds and include integral electrical connections to facilitate simple and fast integration into a control system. The series comprises a range of products classified according to type, size and performance. There are three main sizes: 10 mm, 18 mm and 26 mm. 18 mm has been chosen.

Construction characteristics:

Central body	Extruded aluminium bar with chemical nickel treatment and PTFE (polytetrafleurethylene)
Connection plates	Zincalloy
Operators	Technopolymer
Spool	Aluminium 2011
Piston seals	Oil resistant nitrile rubber-NBR
Spool seals	Oil resistant nitrile rubber-HNBR (Therban)
Springs	Stainless steel AISI 302
Piston	Technopolymer

Adequate lubrication reduces seals wear, just as proper filtering of supply air prevents the build-up of dirt that can cause malfunction.

Operational characteristics:

Fluid	Filtered and lubricated or not lubricated air
Max working pressure	10 bar
Operating temperature	-5° ÷ +50°C
Flow rate at 6 bar with ? p =1	800 Nl/min
Orifice size	Ø 7 mm
Working ports size	G 1/8" – G 1/4"
Switch time	< 15 ms
Weight	175 gr
Maximum dimension	54 x 18 mm and length of 148 mm

The weight of the accessories to fix it has to be added to the weight of the valve. This is about 50 gr.

The 15 mm miniature solenoid valve with 1,1 mm orifice has been selected for piloting this series of valves. This results in low response times and reduced power consumption.
There are different tension to control the coil:
- miniature solenoid 12 VDC
- miniature solenoid 24 VDC
- miniature solenoid 24 VAC
- miniature solenoid 110 VAC
- miniature solenoid 220 VAC

It is sufficient to select signal command in VDC.

Magnetic sensor. The limit switches, or magnetic sensors, have to be mounted on a cylinder with magnetic piston. These, when hit by the magnetic field generated by the piston as it approaches, close the circuit sending an electrical signal by relè solenoid valve control, etc, or converse with the controlling electronic system situated on the machine. Magnetic sensors with ampulla Reed type and Hall effect are available.
For more details see the website.

New cylinder. There is always a continuing evolution of the products, so a new series of cylinder completely in technopolymer (PNEUMAX company) has just been developed. One of the main characteristics is the kind of material used, in fact the head and the body are realized in technopolymer reinforced with glass fiber that confers mechanical characteristics similar to aluminium.
The technopolymer allows to reduce the weight of the cylinder and to increase the movement speed of the piston. Now the bore are available up to Ø 25 mm, but I think in the future the companies will start to produce also the others bores.

Construction characteristics:

Head	Nylon 66 reinforced with glass fiber
Body	Nylon 66 reinforced with glass fiber
Rod	Stainless steel for magnetic cylinder
Piston seal	Special NBR shore self-lubricating
Rod seal	Self-lubricating polyurethane
Clamp	Stainless steel

Technical characteristics:

Fluid	Filtered and preferably lubricated air
Max pressure	8 bar
Working temperature	-5° ÷ +50°C

6.2 HYDRAULIC SYSTEM

The hydraulic system uses hydraulic fluid pumped to high pressure and transmitted by means of the strong pipes to actuators that move things. In the hydraulic systems there are sources of energy, methods for transporting that energy, and then methods for expending that energy. The hydraulic pumps quash the hydraulic fluid, creating fluid at high pressure. They have a power density about ten times greater than an electric motor, and are powered by an engine or electric motor. The theory behind hydraulic equipment is the fluid pressure. A force acting on a small area can create a much larger force by acting on a larger area by virtue of hydrostatic pressure. A large amount of energy can be carried by a small flow of highly pressurized fluid. It is controlled by a control valve and distributed through hoses and tubes. In order for hydraulic fluid to do work it needs to flow to the actuator where it does work, and afterwards return to a reservoir. The fluid is then filtered and re-pumped. Hydraulic cylinders are hydraulic linear actuators driven by pressure differential in the cylinder chambers and they convert hydraulic pressure into linear motion. In the hydraulic circuit, the purpose of the directional control valves is to direct the fluid to the desired actuators.

6.2.1 Hydraulic circuit

The hydraulic circuit is similar to the pneumatic circuit, to activate the cable there is a double action cylinder controlled by an electromagnetic valve. The cylinder has a magnetic switch, the sensor is used to detect the piston’s position inside the cylinder. All these components are inside the dummy, instead the other components that create the energy are outside the dummy (for instance pump, filter, reservoir etc..) and the pressure fluid is carried to valve by means of the pipes (Fig. 6.3).

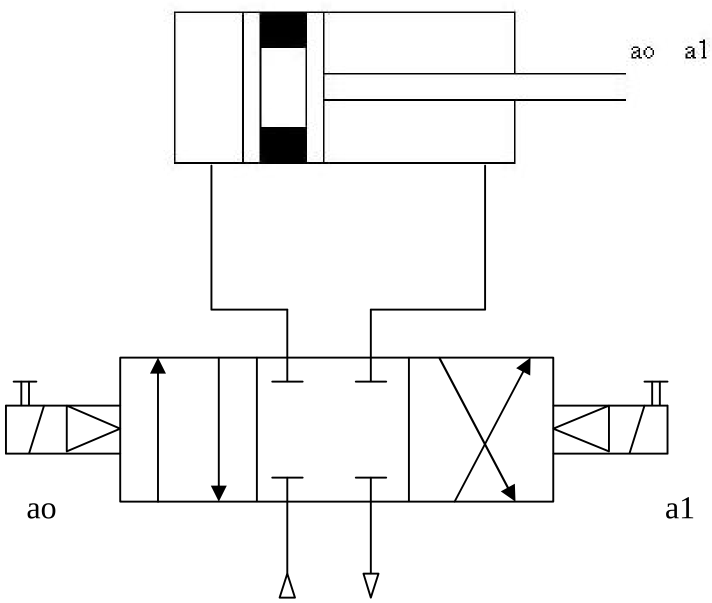


Fig. 6.3 Hydraulic circuit inside the dummy: cylinder with magnetic sensor, electromagnetic valve.

The hydraulic fluid is drawn from a reservoir, through a filter and then is pumped to the directional control valve. If the main valve isn't engaged, the fluid returns to the reservoir, otherwise goes to a hydraulic cylinder. The pump will supply a fluid at high pressure. The fluid that comes back to the reservoir during the movement of the piston will be at low pressure. The magnetic sensor inside the cylinder detects the piston's position and sends an electric signal to the control unit (PC or PLC) that analyses and elaborates the information like in the pneumatic circuit. Afterwards the control unit transmits will send an electric signal that switches the valve.
The representation of this process is the same of the diagram in figure 6.2.

6.2.2 Dimensioning of the hydraulic circuit

The operating fluid pressure in the hydraulic circuit is within the range 100-200 bar, but the exercise normal pressure used is 160 bar.
The formulae used for the dimensioning of the hydraulic cylinder are the same used to dimension the pneumatic cylinder, in the formulae only the value of the pressure, that now is 160 bar, changes.
Using the force for the muscle extensors equal at 980 N (about 100 kg) and the pressure equal 160 bar, the cylinder bore is 9 mm. It is better to choose a bore equivalent to 10 mm because when the piston is pulled, all surface less the surface of the rod have to be considered.

Using the force for the muscle extensors equal to 600 N (about 61 kg) and the pressure equal to160 bar, the cylinder bore is 7 mm. It is better to choose a bore equivalent to 8 mm.

The stroke of the cylinder is like the muscle length, with a maximum stroke equal to 170 mm.
Therefore if the time necessary for a muscle to generate an adequate strength is between 100-170 ms, the speed basic profile is the trapezoidal move and the move time is divided equally into 3 parts. The first part is acceleration, the second constant velocity and the third part deceleration. The peak speed of the cylinder is 1.5 m/s and the acceleration is 18 m/s^2 almost 2g.

6.2.3 Choice of the hydraulic components

In basis of all data previous evaluated, now the best cylinder and valve suitable for the circuit have to be chosen.
The main characteristic of the cylinder and valve are:

1) the cylinder has to have a magnetic sensor that detects the piston's position inside the cylinder, the valve has to be activated electromagnetically. It is important that these two components have this kind of feature because, the sum of feedback signal time, the time of analysis and elaboration data, the time switching of the valve and beginning of the piston moving have to be within the range of the time to reflex (26-92 ms). Therefore only electromagnetic components can have time of answer in the order of the millisecond;
2) the cylinder bore for muscles extensors of 10 mm and for muscles flexors of 8 mm;
3) the cylinder stroke between 70-150 mm, this depends in which position the cylinder will be fixed (stroke is the distance between fully extended and fully retracted rod positions);

4) the peak piston speed is 1,5 m/s. In this case it is important to find a particular system for the piston’s deceleration without crash because the slowing down of the piston is very fast;
5) it is important to consider the weight of the cylinder and of the valve;
6) it is important also to consider the maximum dimension of the valve and the cylinder because there is a limited space in the dummy.

There are a lot of companies producing hydraulic components. I looked in the internet in the web site of the companies but I didn’t find factories that produce cylinders with bore of 10 mm and 8 mm. The smallest cylinder products, has a bore of 20 mm. In the our case I think the compact hydraulic cylinder is useless.
The characteristics of the valves and the hydraulic cylinders are compatible with the characteristics necessary to realize the hydraulic circuit. There is only one big problem, that is the weight. Working with pressure of 160 bar, the cylinders and the valves are made with strong materials like steel and the components are thick otherwise they don’t resist at this pressure. The cylinder weight of bore 20 mm is around 1,5 kg, the correct value of the weight depends from the stroke. The maximum dimension are 44 x 44 mm with length of 80 mm and are reasonable with the available space. The valve weight is more than 1 kg and the maximum dimension are 46 x 81 mm with length of 180 mm.
This problem is greater when one wants to realize a cylinder with bore of 10 mm and 8 mm. I asked some companies if it is possible to make these kind of cylinders. They answered me that it is possible to create them, but the weight is only 30-40 % less of the cylinder weight with bore 20 mm. The problem is given by the high pressure, it is impossible to make the cylinder with light materials and thin structures. Also smaller are the dimensions, more the weight of the screws and brackets used to fix the components affect the total weight.
The companies told me that only one problem can arise, that is to realize the seals inside the cylinder, because they have small sizes and have to resist at the high pressure. Therefore seals with these sizes and characteristics aren’t available. They weren’t sure that seals in polyurethanes are sufficient, but only a real test can say if it is true or not.
In the end of this discussion, there aren’t in commerce these hydraulic components, but it is possible to make them even if probably the weight will be a problem.

6.3 MOTOR LINEAR ACTUATOR

Motors convert electrical energy into mechanical energy. The basic stepper motor creates rotary motion of a magnet rotor core through the use of pulses and electromagnetic field passing around the core. The movement created by each pulse is precise and repeatable, which is why stepper motors are so effective for positioning applications.
A linear actuator converts rotational motion into a linear motion, with the precision depending from the step angle of the rotor and the method chosen to accomplish the conversion. The rotary motion of a stepper motor can be converted into linear motion by several mechanical means. These include rack and pinion, belt and pulleys and other mechanical linkages. All of these options require various external mechanical components. The most effective way to accomplish this conversion is within the motor itself.
Conversion of rotary to linear motion inside a linear actuator is accomplished through a threaded nut and leadscrew. The inside of the rotor is threaded and the shaft is replaced by a lead screw. In order to generate linear motion the lead screw must be prevented from rotating. As the rotor turns the internal threads engage the lead screw, resulting in linear motion. Changing the direction of rotation reverses the direction of linear motion. The basic construction of a linear actuator is illustrated in figure 6.4.

The linear travel per step of the motor is determined by the motor's rotary step angle and the thread pitch of the rotor nut and leadscrew combination. Coarse thread pitches give larger travel per step than fine pitches screws.

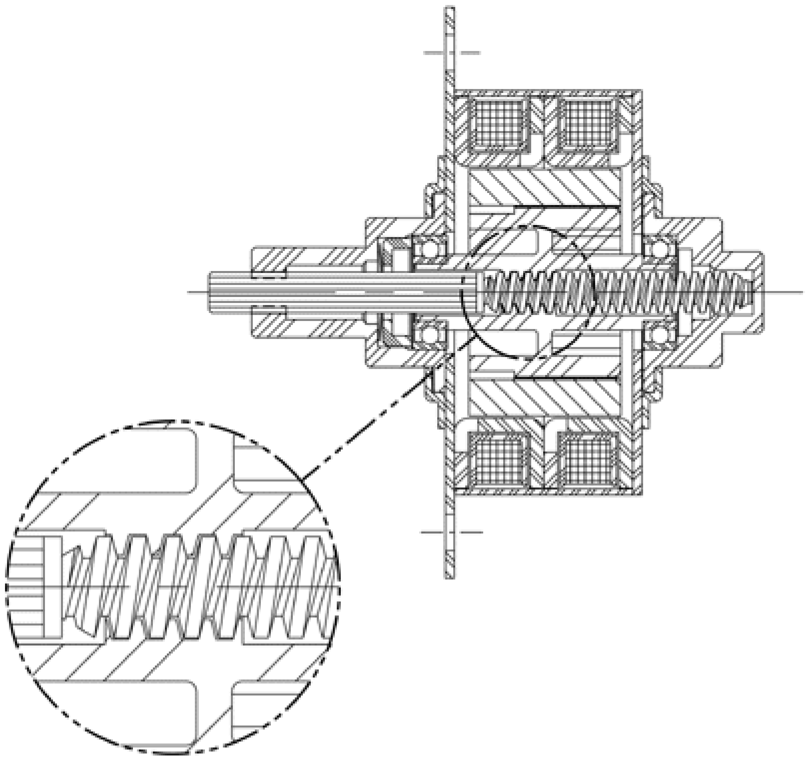


Fig. 6.4 Linear actuator cut away showing threaded rotor to leadscrew interface.

However, for a given step rate, fine pitch screws deliver greater thrust. Fine pitch screws usually can not be manually "backdriven" or translated when the motor is unenergized, whereas many coarse screws can. A small amount of clearance must exist between the rotor and screw threads to provide freedom of movement for efficient operation. This results in .001" to .003" of axial play (also called backlash). If extreme positioning accuracy is required, the backlash can be compensated by always approaching the final position from the same direction.
Accomplishing the conversion of rotary to linear motion inside the rotor greatly simplifies the process of delivering linear motion for many applications. Because the linear actuators is self contained, the requirements for external components such as belts and pulleys are greatly reduced or eliminated. Fewer components make the design process easier, reduce overall system cost and size and improve product reliability.

6.3.1 Circuit linear actuator

The stepper motor allows easy open-loop digital position control. The simplest configuration uses a slave driver which takes step and direction signals from another source (such as a PC, a PLC, or a microcontroller) and supplies the required voltage and current to the motor coils in the proper step sequence (Fig. 6.5).

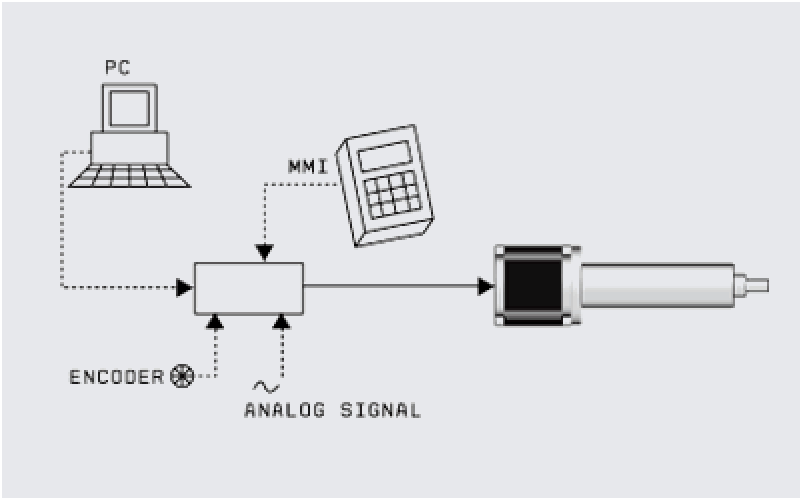


Fig. 6.5 Circuit linear actuator

A slightly more specialized controller allows the operator to control the speed and direction of the actuator with an analog voltage. If more sophisticated control is required, a programmable controller allows to use an external optical encoder (to check the displacement of the actuator). Only the linear actuator is inside the dummy, the other components are all outside and only the electrical cables connect the slave driver with the linear actuator.

6.3.2 Choice of the linear actuator components

The linear actuator characteristics are the same of the pneumatic and hydraulic cylinders, lightness, high acceleration, peak of speed of 1,5 m/s, etc..
I looked in the internet inside a lot of company's web sites but, I didn't find motor linear actuators with all the characteristics required.
In this case it is difficult to put together the weight, the force, high acceleration and speed.
One instance to understand better which are the problems, is to see the two diagrams below that belong to the series of linear actuator of the Haydon company (the diagram of these linear actuator is shown, but the linear actuator of the other companies are more or less the same).
These linear actuators are made with stepper motor, and the conversion of rotary to linear motion inside a linear actuator is accomplished through a threaded nut and leadscrew.

The first linear actuator belongs to series 57000 with dimension of 54 x 54 mm and length of 90 mm and weight of 511 gr (Fig.6.6).
In this diagram it is possible to see:
- if increasing the speed of the stepper motor (number of steps in one second), the thrust decreases;
- if increasing the linear travel per step (the number on the curves), the thrust decreases.

So if the value of the force is chosen, the speed and the linear travel per step are too low and vice versa.

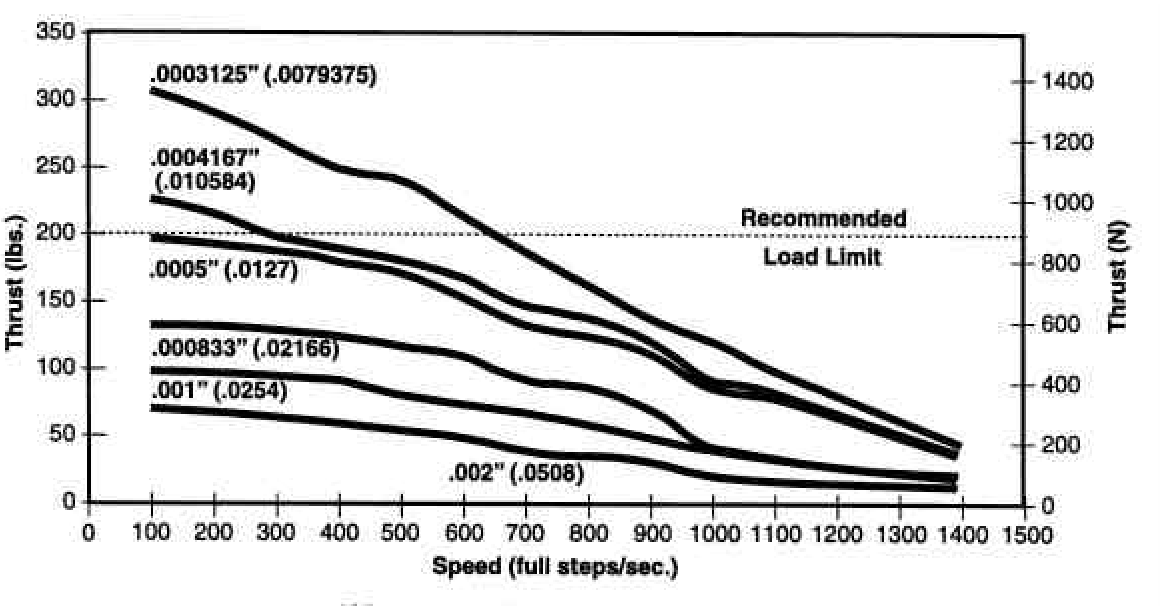


Fig.6.6 Diagram of the thrust force and speed for series 57000. The numbers on the curves indicate the linear travel per step (in inches and millimetres).

The second linear actuator belongs to series 87000 with dimension 87 x 87 mm and length of 120 mm and weight 2,3 kg (Fig.6.7).

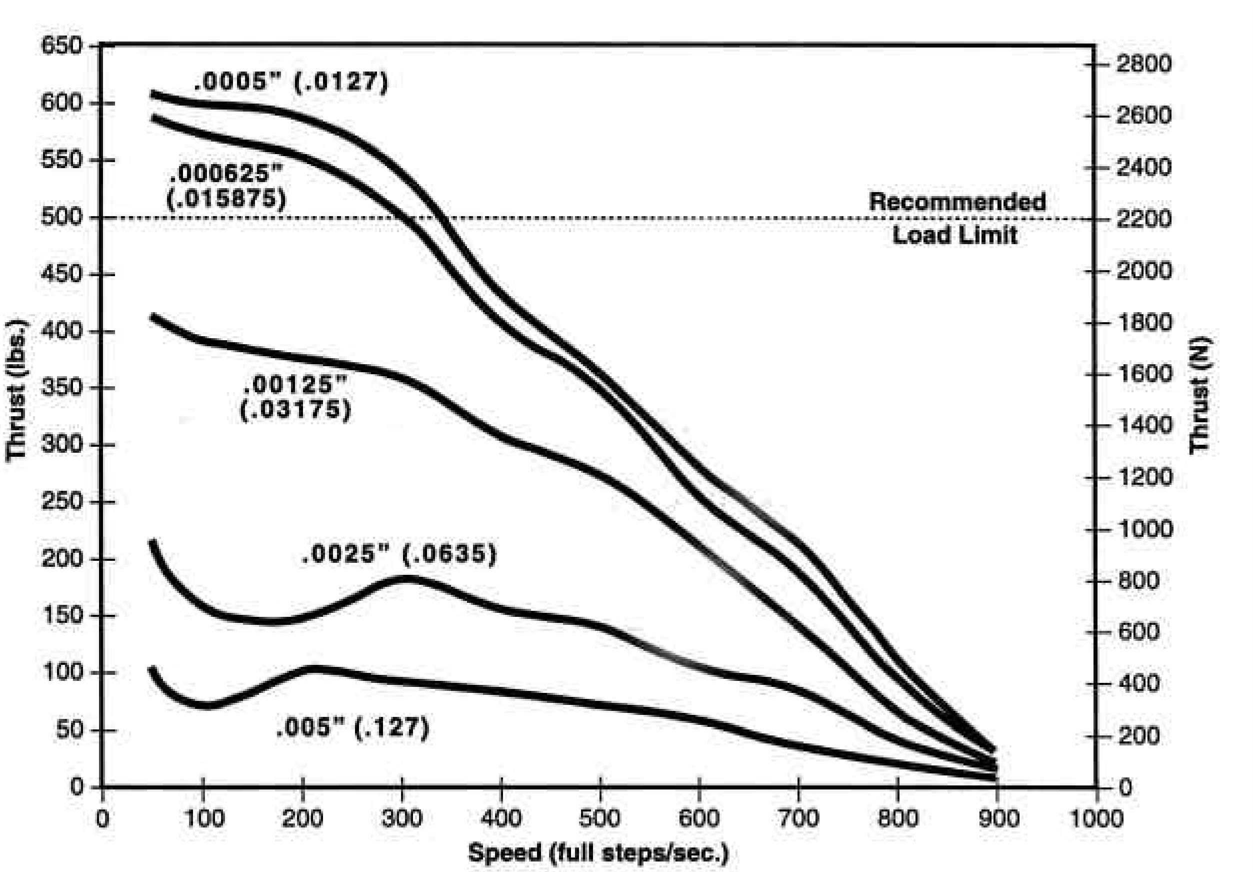


Fig. 6.7 Diagram of the thrust force and speed for series 87000. The numbers on the curves indicate the linear travel per step (in inches and millimetres).

This kind of linear actuator is bigger than the one before, therefore, with the same strength it is possible to work with higher value of speed and linear travel per step, but the problem is due to the weight, because the coil and the system to accomplish the motion conversion are bigger and heavier.
In the end of this discussion, I didn't find available the linear actuator suitable for our requests.

6.4 FLUIDIC MUSCLE

In nature, muscles work perfectly, they contract powerfully and relax in a controlled manner. FESTO company has managed to implement this principle, which applies to the human muscle, in an industrial product. An innovative pneumatic drive is the result: the Fluidic Muscle. This innovation is based on aramid fibers, arranged in rhomboidal fashion and are embedded in a flexible hose. Tremendous tractive force is generated once compressed air is applied (Fig. 6.8).

Fig. 6.8 Picture of the Fluidic Muscle

General characteristics of the Fluidic Muscle are:

1) in the expanded condition it develops up to ten times more power than a conventional pneumatic cylinder of the same diameter;
2) highly dynamic response, even at high loads, acceleration values up to 50 m/s^2 (the lightweight Fluidic Muscle is ideally suited to applications with high acceleration requirements);
3) it produces the same force with only one third of the cross section required by a conventional cylinder;
4) no mechanical parts moving against one another. Thus it works smoothly and enables uniform movement even at low speeds without slip-stick effect;
5) simple positioning controlled by means of the pressure using the simple technology without displacement encoders;
6) the application are as the single-acting actuator or the pneumatic spring.

Mode of operation. Fluidic Muscle is a membrane contraction system or, to put it more simply, a tube which contracts under pressure. It consists of a contraction system and appropriate connectors. The contraction system is formed by a pressure-tight length of rubber hose, sheathed in high-strength fibres. The fibres create a rhomboidal pattern with a three-dimensional grid structure. When internal pressure is applied, the hose expands in its peripheral direction, thus creating a tensile force and a contraction motion in the muscle's longitudinal direction. The usable tensile force is at its maximum at the start of the contraction, and then decreases in a virtually linear manner as a function of the stroke. An efficient operating range is provided with contractions of up to 15 % of the nominal length (Fig. 6.9).

Fig. 6.9 Partial section of the Fluidic muscle

There are two kind of Fluidic Muscle:

1) Fluidic Muscle DMSP, with press-fitted connections;
2) Fluidic Muscle MAS, with screwed connections.

The Fluidic Muscle DMSP with press-fitted connections is the results of a thorough analysis of the requirement's specification that already existed for the Fluidic Muscle MAS. Therefore the Fluidic Muscle DMSP is considerably lighter (up to 30% less weight), more compact (25% more compact cross section) and more durable than Fluidic Muscle MAS.

Below the characteristics of one Fluidic muscle DMSP size 20 are shown, but there are different sizes: 10, 20 and 40 for the Fluidic Muscle DMSP with press-fitted connections and also for the Fluidic Muscle MAS with screwed connections (for more details to see the FESTO web site).

General technical data:

Size	20
Pneumatic connection	G ¼”
Design	Contraction membrane
Mode of operation	Single-acting, pulling
Internal diameter	20 mm
Nominal length	From 60 to 9000 mm
Force at max. permissible operating pressure	1500 N
Max. additional load, freely suspended	80 kg
Max. permissible pretensionsing	4% of nominal length
Max. permissible contraction	25% of nominal length

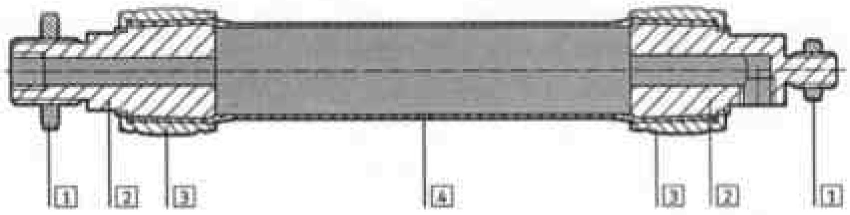
Max. hysteresis	= 2,5% of nominal length
Max. relaxation	= 3% of nominal length
Repetition accuracy	= 1% of nominal length

Operating and environmental conditions:

Operating pressure	0 ÷ 6 bar
Operating air	Filtered compressed air, lubricated or unlubricated
Ambient temperature	-5 ÷ + 60 C°

The basic weight at 0 m length for Fluidic Muscle DMSP with press-fitted connection and size equal 20 is from 169 to 222 [gr] and, each additional weight per 1 m length is 178 [gr].

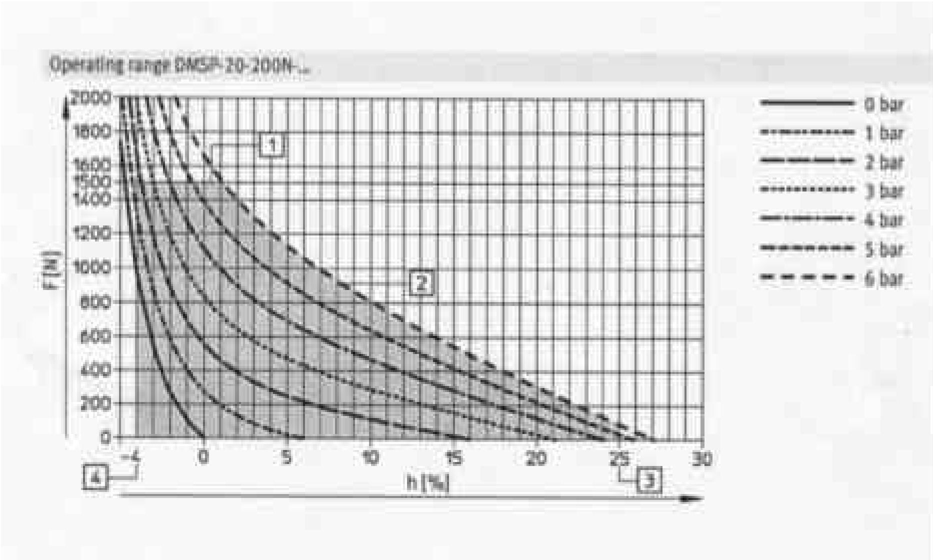
Materials and section view of the DMSP



1. Nut	Galvanised steel
2. Flange	Wrought aluminium alloy, clear anodised
3. Sleeve	Wrought aluminium alloy, clear anodised
4. Membrane	Chloroprene, aramide

Diagram of the operating range for DMSP (size 20)

1. the upper limit of the grey area describes the minimum theoretical force at maximum operating pressure;
2. the right limiting curve of the grey area describes the maximum permissible operating pressure;
3. the right vertical limit of the grey area describes the maximum permissible contraction;
4. the left limit of the grey area describes the load limit of the muscle defined by the maximum pretensioning.



This diagram is for the Fluidic Muscle DMSP size 20. It is possible to see that with a strength of 980 N (like the extensors force) the Fluidic Muscle has a contraction around 7% and with strength of 600 N (like the flexors force) it has a contraction around 14%. If it wants to increase this value of contraction it is possible to choose the Fluidic Muscle DMSP size 40, with a value of force contraction of 25%, but it is bigger and heavier (the weight is about 600 gr). Otherwise with this system it is easy to put more cables to reproduce the human muscle system, so the value of strength decreases (the force value of 980 and 600 N were for three extensors and two flexor) and, the percentage of contraction increases.

7. CONCLUSION AND RECOMMENDATIONS

This thesis is divided in two big parts. The purpose of the first part is to study a background of the most important properties and characteristics of the human muscle system, and analysis of the mathematical modelling to understand the behaviour of the neck in different types of impacts and speeds. The second part is to make components that replicate the muscle system, and in the end to think about the different possible solutions to move the cables with human neck motion.

The most important properties and characteristics of the human muscle system found are:

1) time to reflex, the value of this time is within a range of 26-92 ms for the unexpected impact;
2) the time of muscle contraction to reach maximum force is in the range 115-251 ms after the onset of acceleration. This means that the time necessary for a muscle to generate a adequate strength is estimated between 70-115 ms until a maximum of 150-170 ms;
3) passive and active muscle behaviour. The muscle is only able to contract and is not able to stretch alone, consequently, for instance, a movement of flexion-extension antagonist muscle are activated;
4) the force-deformation velocity relationship. The dynamic force in the cervical muscles may be around 40-50% higher than that generated under static condition. It is very difficult to choose exactly values of muscle strength to plan the prototype of the muscle system. In literature there are different values of muscle cross sections, torques around head gravity centre, and around occipital condyles. The reason is that the moment arms and the cross sections are very difficult to determine in vivo. Even if advanced methods are used such as computer tomography, there are large variations of literature data because they depend from the method utilized to analyse, the kind of people analysed, if more trained or not, the age, the weight, the height.
5) the weight of the human chest is around 15 kg;
6) the range of change of the muscle length is 70-150 mm. This value of range has been calculated considering the results of the head gravity center angular displacement obtained in a mathematical modelling. The range depends from the severity of impact pulse, different reflex time, initial muscle tension and other parameters;
7) the cross section of the stronger and more superficial cervical muscles were chosen (see chapter 3), because the deep cervical muscles have only stability function of the head.

The muscle system has a different behaviour in the different kind of impacts, speed and strength.
The muscles can significantly affect kinematics of the head-neck complex in 15 g impact when the following conditions are satisfied simultaneously:
1) peak of T_1 horizontal acceleration is less than 70 g;
2) reflex time is lower than 60 (80) ms.
When these conditions were satisfied, the maximum angle of the head flexion was decreased by muscle tension up to 40%. When the peak of T_1 acceleration was about 70 g, muscles were not able to significantly influence kinematics of the head-neck complex, even for very low values of the reflex time. It can be concluded that the muscle effect is negligible in impacts which are more severe than 15 g.
The kinematics of the head-neck complex depends also from the initial tension muscles. The pre-tense activity of the muscles decreases the maximum angle of the head flexion.

In the second part of the thesis, different components have been planned and they can be inserted in the new prototype. One component is the interface between the head and occiput where the cables are attached, and the second component is a piece to give the bending to the cables, for a better replica of the muscle system. They are made in acetal plastic because this material has low weight but is very strong and at the same time it is easy to shape.
Cables have been thought to reproduce the muscle system. They have been made in steel because it has a high value of the Young's modulus.

This thesis is only a theoretical study and can be used to give some ideas about how to replicate the human muscle system. Therefore each component that has been thought, can be modified. Also the number and the position of the cables can be modified, because the real test or studies with dynamic program like ADAMS or MADYMO haven't been done. This kind of analysis is necessary to understand the better stability of the head in flexion-extension or in lateral movements.

Pneumatic, hydraulic and motor linear actuators are the systems analyzed to move the cables with characteristics and properties similar to the human muscle system. Having compared the three different systems, I think that the pneumatic circuit is the best to replicate the movement of the muscle system. The valves and cylinders already existing are able to reproduce the human parameters. If the cylinder in technopolymer will be produced (if it isn't produced, I think, it is possible to make one special cylinder with this material), it will improve the characteristics of speed and movement of the piston and decrease the cylinder weight.
It is possible to use the hydraulic circuit because the speed of the piston and the different time of movement are within the maximum value range, but probably, the components will have problems of weight.
I don't think it is possible to use the motor linear actuator because it is difficult to find the motor linear actuator with the all characteristics requested. Maybe there are some special applications, that use particular motor linear actuators, useful for our purpose, but I didn't find them.

The Fluidic Muscle is a very interesting system to replicate the human muscle system because it is already thought to work like the human muscle. It has high acceleration, high force, small dimension, is light, there aren't components in movement and the position in controlled by means of the pressure.
There is only one problem. The Fluidic muscle is able to contract his length until 25% of nominal length, but is only able to extend his length until 4% of nominal length. When the flexors are contracted, the extensor are elongated, more or less with the same length. Therefore it is necessary to think a system to connect with the Fluidic Muscle to supply the elongation. Or, maybe, it is possible to give a small pressure, for instance 3 bar (that is a nominal length), so the Fluidic Muscle has a pre-elongation and after, if it needs to contract, it can provide more pressure, otherwise, if it needs to lengthen, it can take off the pressure.
Unfortunately I have discovered the Fluidic Muscle only in the end of my work, so I didn't have a lot of time to study in depth this system.

In the end I think the pneumatic method is the best solution compared with the hydraulic and magnetic system, but the Fluidic Muscle, if studied in depth, I think will be the best solution.

If a pneumatic solution will be chosen, the future work is to reproduce in reality this system and to do the real test, because the different times of answer of the valve and movement of the piston can be exactly evaluated only in experimental tests. If the Fluidic Muscle is chosen, the future work is to study in depth this system, to evaluate the system to connect it with the

Fluidic Muscle and to provide the elongation or to try with a pre-elongation (there is on the FESTO web site a software to help the people to determinate the correct size of the Fluidic Muscle).

This work can be considered a good basis for future studies in the realization of the dummy neck muscle system.

LIST OF FIRMS

Pneumatic web site:

www.generalmatic.com

www.pneumaxspa.com

www.metalwork.it

www.festo.com

www.mecfluid2.it

www.azpneumatica.com

Hydraulic web site:

www.parker.com

www.duplomatic.com

www.dogican.it

www.bierihydraulik.com

www.generlmatic.com

Motor linear actuator web site:

www.hsi-inc.com

www.skf.com

www.ultramotion.com

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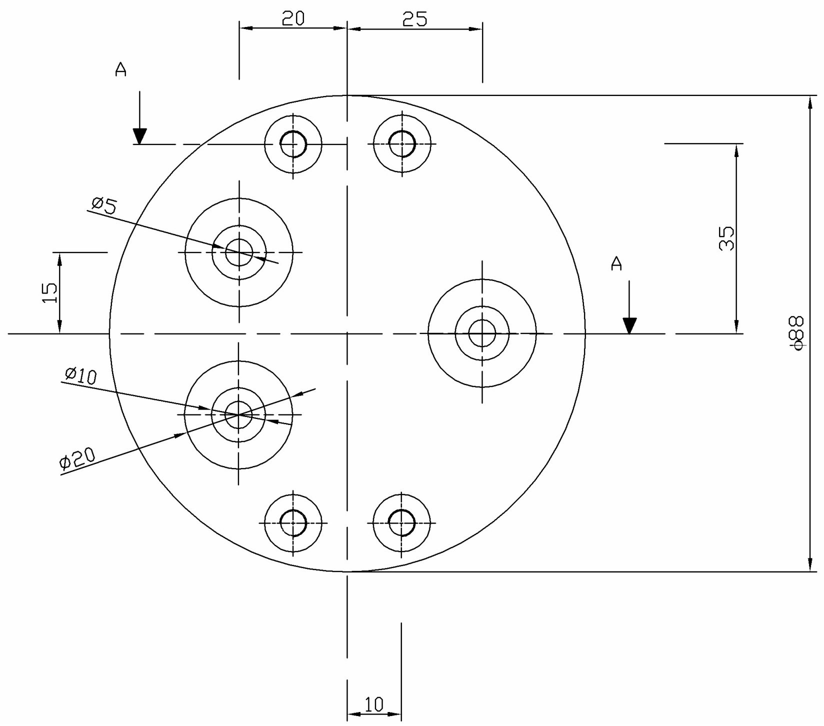
APPENDIX

NEW COMPONENTS:

1) Interface between the head and occiput where the cables are linked.

Superior view.



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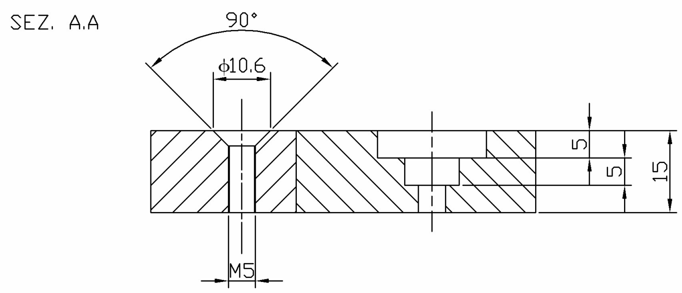
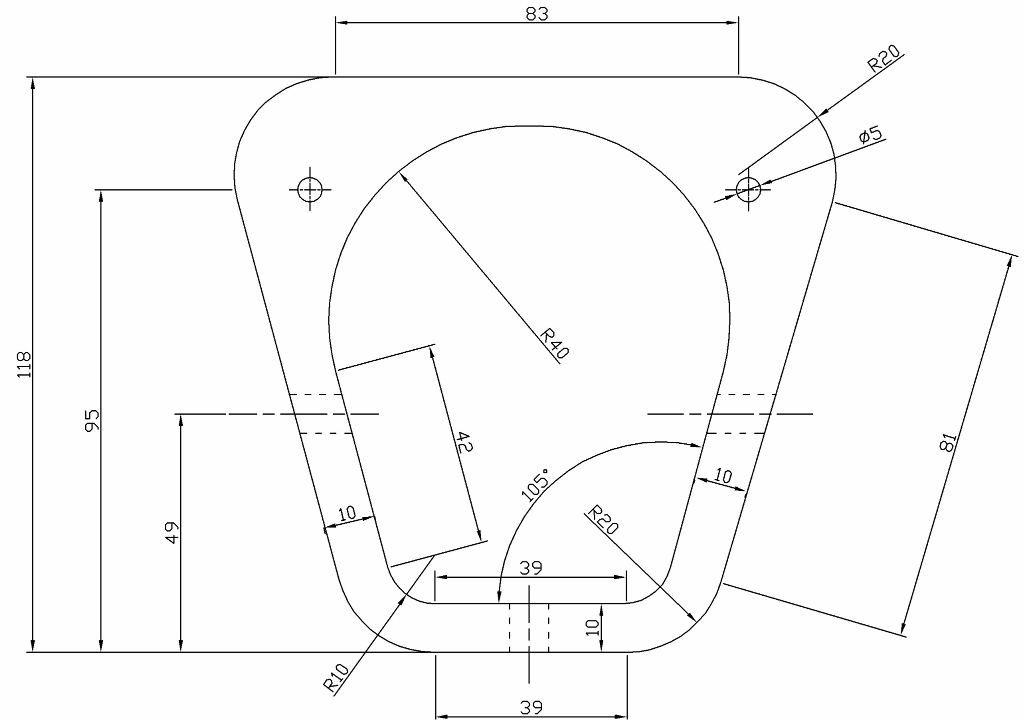
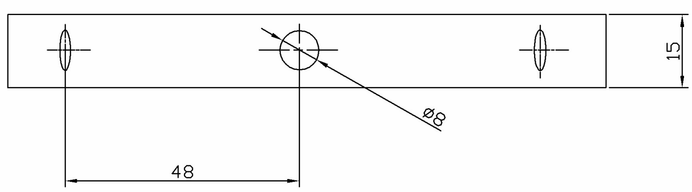


2) Piece that gives the curvature at the cables.

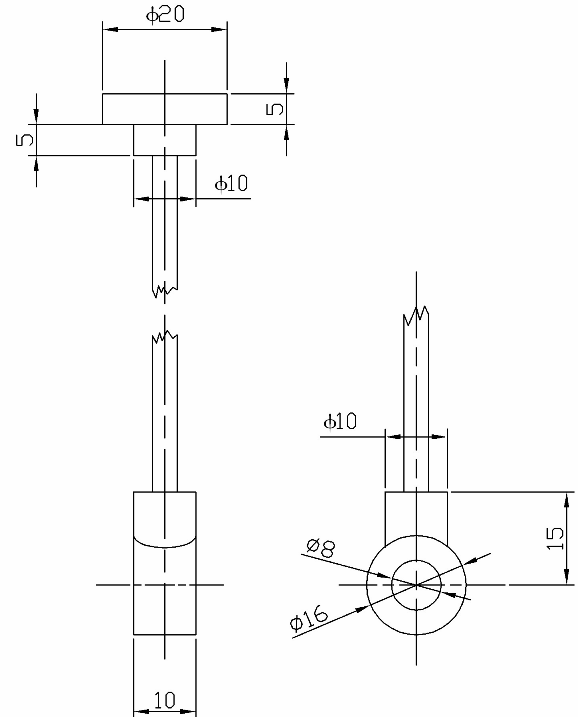
Superior view.



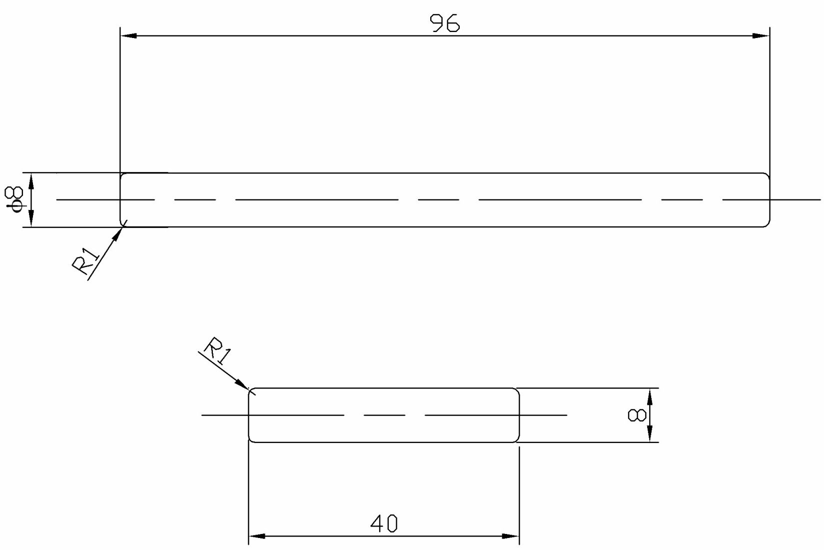
Frontal view.



3) Cable.



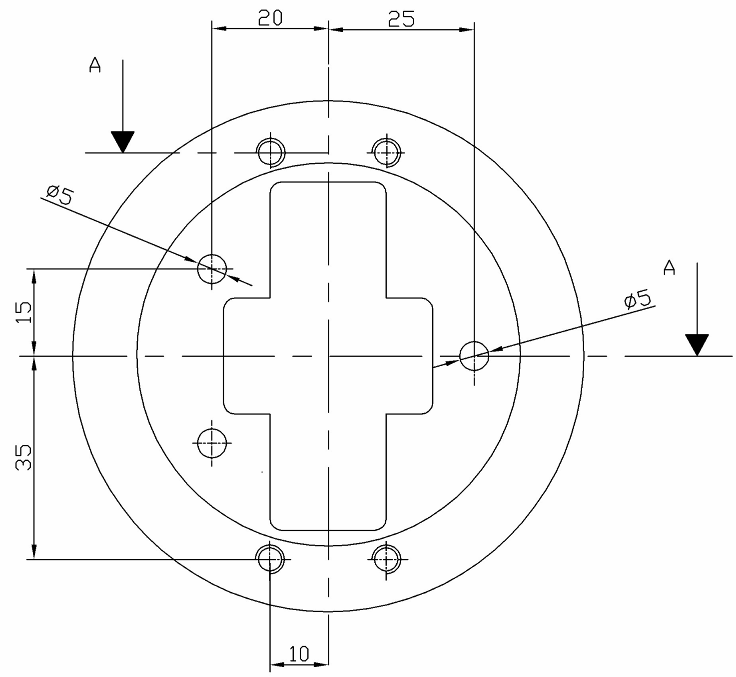
4) Pins.



MODIFICATION AT THE PIECES ALREADY EXISTING.

1) Occiput interface.

Superior view.



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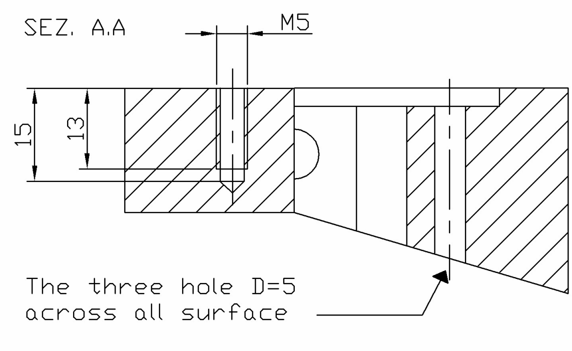
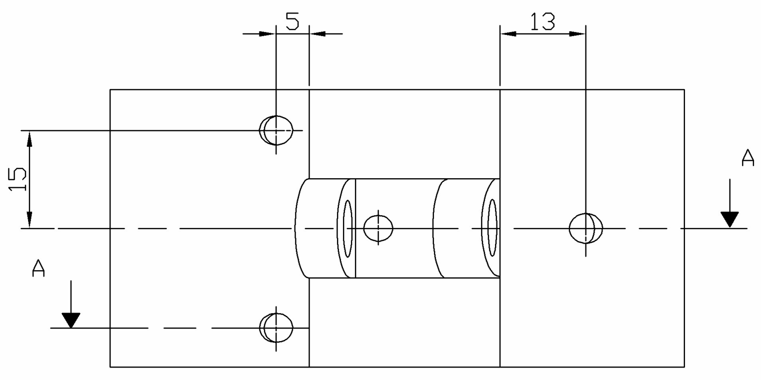


2) C_1 vertebrae.

Superior view.



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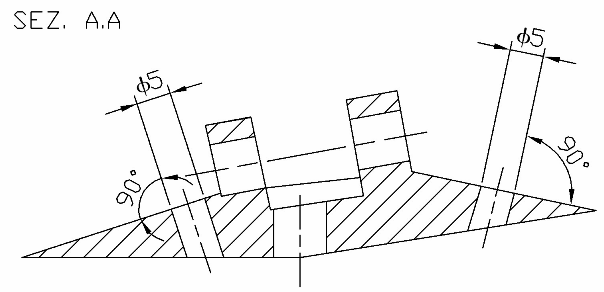
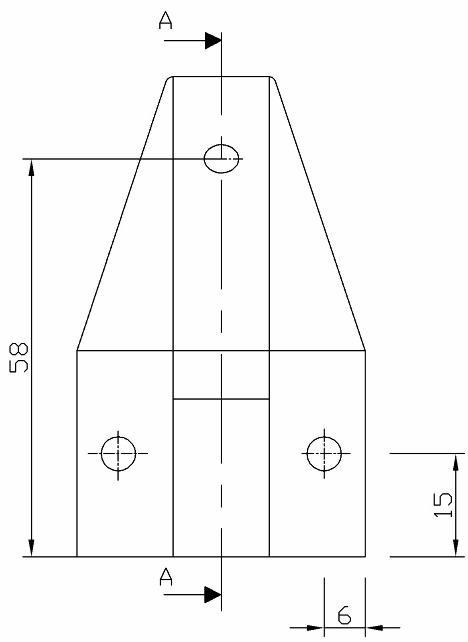


2)Lower cervical vertebrae.

Superior view.



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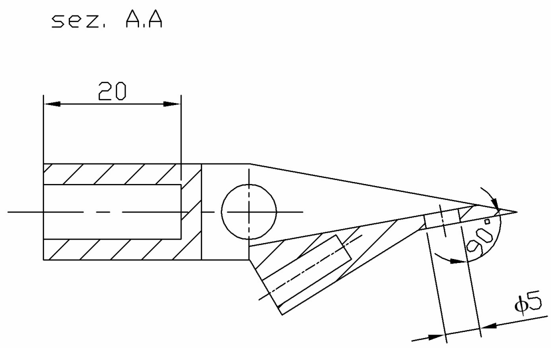

Frontal view.