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Deep muscle function in the cervical spine: Application to musculoskeletal modelling

Jonathon Green
Edith Cowan University

EDITH COWAN UNIVERSITY

School of Biomedical and Sports Science

Deep Muscle Function in the Cervical Spine: Application to Musculoskeletal Modelling

Honours Thesis

Student's Name: Jonathon Green

Student Number: [redacted]

Principal Supervisor: Dr Angus Burnett

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USE OF THESIS

DECLARATION

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ABSTRACT

Neck pain in occupational tasks is a problem that involves a high cost to society therefore, understanding the mechanics of musculoskeletal loading is important in formulating preventative rehabilitation strategies. Musculoskeletal modelling of the neck provides a means by which to calculate loads on the components and muscles of the neck thus allowing quantitative data to be gained non-invasively for a wide range of occupational tasks. Anatomically detailed, electromyography (EMG) driven neck models require deep muscle EMG activation profiles which are difficult to attain without invasive EMG procedures. The aim of this study was to determine whether EMG activity of semispinalis capitis (a posterior deep neck muscle) could be predicted from the EMG activity of trapezius (a posterior superficial neck muscle) and whether EMG activity from splenius capitis (a posterolateral deep neck muscle) could be predicted from the EMG activity of levator scapulae (a representative posterolateral superficial muscle). Surface EMG was recorded unilaterally from two sites around the neck at the C4/5 level and intramuscular EMG was recorded using fine-wire EMG electrodes on six healthy male subjects. Subjects performed a series of maximal and sub-maximal isometric contractions against the torque arm of an isokinetic dynamometer in the direction of extension and right lateral bending with the head in neutral and non-neutral (20° flexion, 35° extension and 30° lateral bending) postures. EMG data were normalised using maximum voluntary isometric contractions (MVIC's) based on a reliability study that was also conducted. The root mean of the squared differences (RMS difference) was used to compare the surface and intramuscular EMG waveforms. RMS difference was chosen as the best indication of predicability, as a quantitative assessment of amplitude difference was obtained using this method. The mean of the RMS difference values between surface and deep musculature in this study was 19.8 %MVIC for the posterior aspect of the neck and 23.9 %MVIC for the posterolateral aspect of the neck. Due to the magnitude of difference between the surface and intramuscular EMG electrodes, it was concluded that EMG activity of the surface muscles does not represent activity of deep musculature in the posterior and posterolateral aspects of the neck.

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CHAPTER ONE

1.0 INTRODUCTION

1.1 Background

The neck provides both stability and movement for the head which contain the body's key sensory organs. The weight of the head (approximately 7.3% of body weight) and its position at the end of the kinetic chain places great strain on the spinal structure and musculature of the neck when various movements are performed. The neck is susceptible to injury in many different situations with rear impact automobile accidents, (Brault, Siegmund & Wheeler, 2000; Magnusson et al., 1999; Svensson et al., 2000; Wittek, Ono, Kajzer, Ortengren & Inami, 2001) aviation (Hamalainen & Vanharanta 1992; Oksa, Hamalainen, Rissanen, Myllyniemi & Kuronen, 1996) and sports trauma (Scher, 1998) being common causes of injury. Furthermore, a number of ergonomic studies have reported a high prevalence of musculoskeletal discomfort in the neck in other occupational tasks. These include dental work, static work at workstations, sitting in front of video display units, sewing machine operation and computer aided designing (Sommerich, Joines, Hermans & Moon, 2000). Research pertaining to musculoskeletal disorders in children and young adults associated with laptop and workstation use is also an important developmental health and ergonomic issue (Harris & Straker, 2000; Straker & Mekhora, 2000; Zandvliet & Straker, 2001).

Calculating loads on the spine and its component structures is a necessary step towards understanding the mechanics of injury (Kingma et al., 2001; Marras et al., 1995; McGill & Norman, 1986; Shirazi-Adl, Ahmed & Shirvastava, 1986). To formulate preventative and best practice rehabilitation strategies, it is vital that the mechanisms of these injuries and the musculoskeletal loading involved be thoroughly examined and understood. Consequently, it has been suggested that the single most important area of study in human movement is the development of accurate, non-invasive means of predicting individual muscle and ligament force-time histories in various body movements

(Dowling, 1997; McGill & Norman, 1986). Measuring muscle forces and joint loads in an *in-vivo* model (e.g. Butterman, Schendel, Kahmann, Lewis & Bradford, 1991) is not widely practised and has mainly been restricted to animal studies due to ethical reasons.

Moroney, Schultz and Miller (1988) developed a musculoskeletal neck model using an optimisation technique in combination with an inverse dynamics approach to measure individual neck muscle loads. The major shortcoming of this technique was that it underestimated joint moments by assuming minimal antagonistic muscle forces. Electromyography (EMG) and EMG-assisted optimisation techniques, which utilised muscular activation levels from both agonist and antagonist muscle groups across joints, to predict joint loads have been developed to overcome this problem (e.g. Choi & Vanderby, 1999; Cholewicki, McGill & Norman, 1995; McGill & Norman, 1986).

With the recent advances in computer technology and animation, graphically based three-dimensional models of structures within the human body have been, or are being, developed. Vasavada, Li and Delp (1998) developed an anatomically detailed model of the neck using a commercially available software package called Software for Interactive Musculoskeletal Modelling (SIMM). This method of modelling is becoming increasingly popular amongst researchers in biomechanics. Muscle morphometric data for this model was determined from cadaver studies by Kamibayashi and Richmond (1998). The musculoskeletal model is typically driven by the EMG activity of the large superficial muscles of the neck however, there are deeper muscles of the neck whose activity also needs to be considered as these have the potential to add to the moment generating capacity of the neck. This problem has also been considered by McGill, Juker and Kropf (1996) for the musculoskeletal modelling of the lumbar spine (McGill & Norman, 1986).

1.2 Significance of the Study

Neck pain in occupational tasks is a problem that involves a high cost to society. With the data collected from this research the validity of a musculoskeletal model may be improved. Therefore, quantitative data on neck loading may be gained non-invasively for

a wide range of occupational tasks. To the author's knowledge no study has examined whether deep muscle activations can be predicted from surface EMG electrodes in the neck, except for Mayoux-Benhamou, Revel and Vallee (1995) who investigated splenius capitis. However, the authors did not analyse results with a precise, quantitative application as performed in McGill and co-worker's (1996) study of the trunk.

1.3 Purpose of the Study

The aim of this study was to determine whether the magnitude of EMG activity of the deep muscles can be predicted from the magnitude of EMG activity of the surface musculature at the posterior and posterolateral aspects of the neck utilising data gained from investigating reliability and normalisation procedures.

1.4 Limitations of the Study

1.4.1 Limitations

i) Surface electrodes do not record differences between parts of a muscle as they record activity of a volume of muscles under the electrodes.

ii) Fine-wire electrodes may not represent activity of the entire muscle.

iii) Noise may be present in data resulting from such events as crosstalk, artefact from heart beat or from swallowing.

iv) Anterior and anterolateral locations of the neck could not be examined due to the difficulty in using intramuscular electrodes in this region.

1.4.2 Delimitations

i) Subjects will be between the ages of 18 and 35 years, and the data emanating from this study will be representative of that age group.

ii) It is assumed that activation from the sampled muscles is symmetrical.

iii) The findings in this study can only be generalised to the posterior and posterolateral aspect of the neck and the muscles under investigation in this study.

1.5 Research Questions

The research questions of this study were: -

i) Can muscle activity from semispinalis capitis (a representative posterior deep muscle) be predicted from muscle activity of trapezius capitis (a representative posterior superficial muscle)?

ii) Can muscle activity from splenius capitis (a representative posterolateral deep muscle) be predicted from muscle activity of levator scapulae capitis (a representative posterolateral superficial muscle)?

1.6 Hypotheses

The hypotheses proposed for this study were: -

i) Muscle activity of semispinalis capitis cannot be predicted from muscle activity of trapezius.

ii) Muscle activity of splenius capitis cannot be predicted from muscle activity of levator scapulae.

1.7 Definitions of Selected Terms

Definitions of terms and acronyms used throughout this paper are provided below: -

i) 60%-MVIC: A contraction performed and sustained at 60% of the determined maximal voluntary isometric contraction.

ii) ID: Isokinetic dynamometer.

iii) C1, C2...C7: Cervical vertebrae 1 to 7.

iv) C4/5: Motion segment formed by vertebra C4 and vertebra C5.

v) CT: Computer tomography.

vi) EMG: Electromyography.

vii) MRI: Magnetic resonance imaging.

viii) MVC: Maximal voluntary contraction.

ix) MVIC: Maximal voluntary isometric contraction.

x) PCSA: Physiological cross sectional area.

xi) RMS: Root mean square (mathematical equation)

xii) SIMM: Software for interactive musculoskeletal modelling (Software package by MusculoGraphics™).

xiii) SNR: Signal to noise ratio.

xiv) SPL: Splenius capitis.

xv) SSC: Semispinalis capitis.

xvi) Sub-MVIC: Sub-maximal voluntary isometric contraction.

CHAPTER TWO

2.0 REVIEW OF LITERATURE

Topics discussed in this review of literature include a brief description of the anatomy of the cervical spine and common mechanisms of injury; an overview of the studies involving EMG of the neck and a brief discussion of normalisation procedures.

2.1 <u>The Anatomy of the Cervical Spine and Common Injuries</u>

Seven vertebrae compose the adult cervical spine, with the most superior two vertebrae known as the atlas (C1) and the axis (C2), and the next five following inferiorly (C3-C7). A flexible column is formed by the connection of the multiple joints allowing the head to move in extension, flexion and lateral bending (Sommerich et al., 2000) thus making three-dimensional relative rotation possible (Panjabi & White, 1990). The head, which rests on top of the cervical spinal column, has a mass approximately 7.3% of an individual's body and exacerbates the loads and stresses experienced by the cervical spine. Active and passive tissue surrounds the spinal column, including muscles that provide stability, generate movement and absorb shocks, loads and stresses (Winters & Peles, 1990).

Cervical disorders are a common problem and it has been reported that more complaints are received due to neck disorders than back disorders (Ylinen & Ruuska, 1994). Injuries to the cervical region possibly arise due to the architecture of the head-neck system and the most common complaint among cervical spine disorders is neck pain however, the exact mechanical cause remains elusive (Bicer, Yazici, Camdeviren & Erdogan, 2004). In a Canadian study, 54% of workers had experienced neck pain within a six month period (Ylinen et al., 2003) and a Finnish health survey determined that 15% of men and 6% of women suffer from chronic neck pain (Ylinen & Ruuska, 1994). In 1995, three million acute whiplash cases were recorded in the USA alone (Brault et al., 2000). Worldwide it has been estimated that 40% of the population suffers, or has suffered from, neck pain of

some description (Sommerich et al., 2000). Whiplash is the most widely studied head and neck injury, commonly occurring from rear-impact automobile accidents and victims may present with chronic symptoms post-injury (Brault et al., 2000). Participants of heavy or physically demanding activities have been associated with cervical spondylosis and degenerative changes, whilst tension neck syndrome or myalgia is commonly present in workers who adopt static postures for extended periods of time, such as dental and computer workers (Sommerich et al., 2000). Fighter pilots are also subjected to high loads, for example the cervical erector spinae muscles exhibited the greatest strain during twisted, rotation-plus-extension positions of the head (Hamalanien & Vanharanta, 1992). This study concluded that the neck muscles' potential to protect the cervical structures against injury was lowest during twisted head positions, when in fact this is the position where it is needed the most.

2.2 An Overview of the Methodologies Used in Cervical Muscle Function Studies

From reviewing the pertinent literature, it is clear that there have been a number of different methods used to attain muscle activation data from the cervical region. This section outlines the typical subjects utilized, descriptions of surface and intramuscular electrodes and a summary of muscle placements used to obtain data from the muscles which are under investigation, or alternatively, were considered for investigation in this study. Table 1 shows typical electrode placements used by previous researchers when investigating the cervical region. The superficial and deep muscles located at the posterior and posterolateral aspects of the neck that will be investigated in this study are trapezius (posterior, superficial), semispinalis capitis (posterior, deep), levator scapulae (posterolateral, superficial) and splenius capitis (posterolateral, deep). Figure 1 shows a magnetic resonance imaging (MRI) scan of the posterior and posterolateral musculature outlining the location of the abovementioned muscles at the C4/5 level. Sternocleidomastoid is also outlined due to its involvement in issues involving electrode placement.

Table 1. Summary of electrode positioning used in previous studies (adapted from Sommerich et al., 2000).

	Superficial Muscles		Deep Muscles	
Author	Trapezius	Levator Scapulae	Splenius Capitis	Semispinalis Capitis
Keshner et al. (1989)	(Surface) Palpated muscle belly around C6/7 level and posterior to insertion on lateral third of clavicle (Fine-wire) Bipolar needle electrode inserted close to the midline (within 2 cm) at C1/2 level		(Surface) 6-8cm lateral and 6cm superior to the bony protuberance at C7 level	(Fine-wire) Inserted 2cm lateral to the midline at C1/2 level
Choi and Vanderby (1999)	(Surface) Placed electrodes at the C4/5 level			
Mayoux-Benhamou et al. (1995)	(Surface) Placed electrodes over the belly of trapezius at the C4 level		(Fine-wire) bipolar electrodes horizontally inserted 2cm from midline at C4 level, at depth 1.45-2.6cm (median = 2.26cm)	
Quessier et al. (1994)		(Surface) Placed between the posterior of SCM and the anterior margin of TRAP, pars descendens	(Surface) 6-8cm lateral and 6cm superior to the bony protuberance at C7 level	(Fine-wire) Inserted 2cm lateral to the midline at C1/2 level
Schüldt and Harms-Ringdahl (1988)		(Surface) Placed between the posterior of SCM and the anterior margin of TRAP, pars descendens		
Magnusson et al. (1999)		(Fine-wire) Electrode at the C4 level, 4 cm from the midline at incline of 20 deg toward sagittal plane	(Fine-wire) Inserted 26mm lateral to C3 at a depth of 27mm	(Fine-wire) Inserted 21mm lateral to C3 at a depth of 34mm
Kramer et al. (2003)				(Fine-wire) Regression equations determined predicted puncture depth = circ*0.178–3.65, predicted puncture angle = circ*0.8+11.65

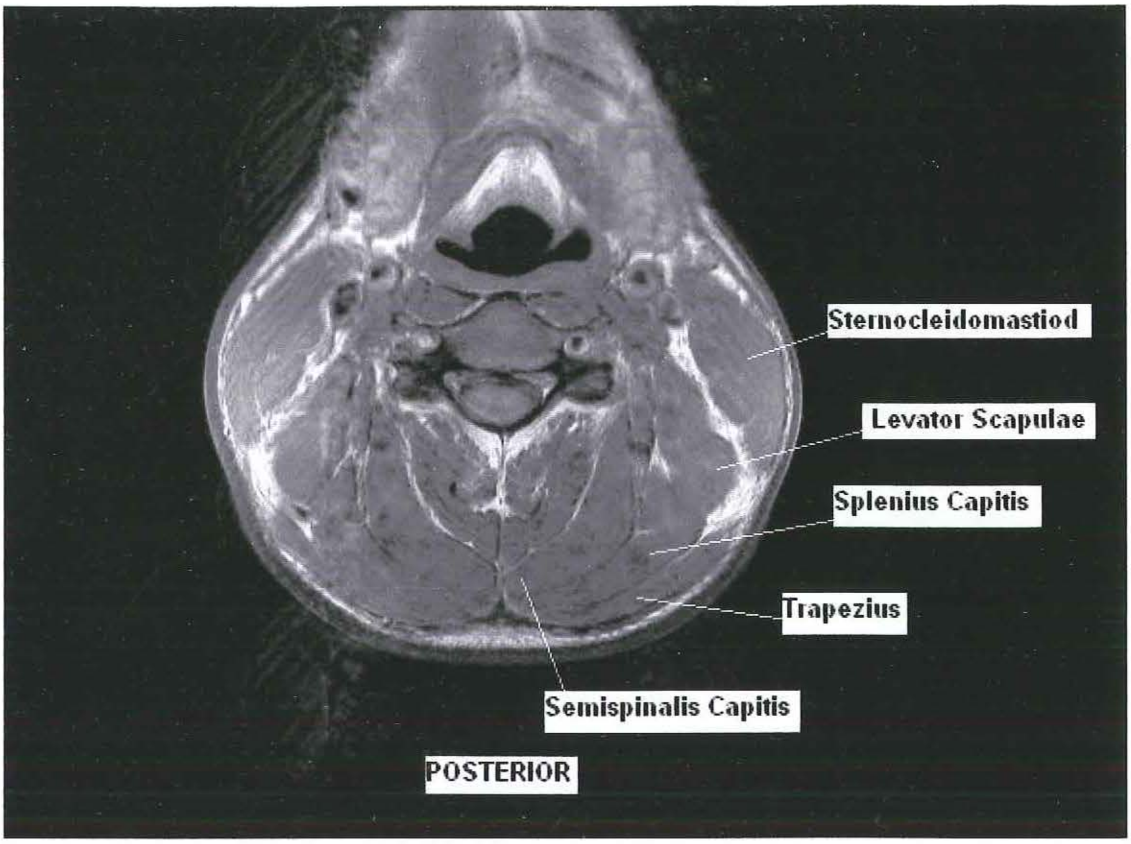

Figure 1: MRI of the posterior and posterolateral musculature of the neck the C4/5 level.

2.2.1 Typical Subjects

Cervical muscle function studies have been conducted with both surface and intramuscular electrodes. Intramuscular electrodes (fine-wire and needle electrodes) have been used to understand the function of neck muscles in both male and female studies. Mayoux-Benhamou et al. (1995) examined a group of subjects (both genders) aged 29-41 years and Keshner, Campbell, Katz and Peterson (1989) utilised a group aged between 22 and 44 years (unspecified gender). Magnusson et al. (1999) used only males aged 24-56 years, with an average age of 40.4 years while Dressler (2000) examined older patients (average age of 54.7 years) of mixed gender with cervical dystonia. Wittek et al. (2001) tested male volunteers only and x-ray was used to confirm that no subjects exhibited degenerative changes in the cervical spine. Neck pain is a common exclusion criterion in

studies involving the use of intramuscular electrodes in the cervical spine (Keshner et al., 1989; Magnusson et al., 1999; Mayoux-Benhamou et al., 1995; Wittek et al., 2001) and other exclusion criteria for EMG studies of the neck exist (Sommerich et al., 2000)

2.2.2 Use of Surface Electrodes

Sommerich et al. (2000) stated that concerns have been raised over the use of surface electrodes to examine the function of splenius capitis and semispinalis capitis as crosstalk (the detection of unwanted activity from adjacent muscles) from trapezius, levator scapulae and sternocleidomastoid may lead to questionable results. Keshner et al. (1989) reported that splenius capitis was involved in both the anterolateral stabilisation and posterolateral stabilisation of the neck. These findings could have been disputed due to crosstalk as Mayoux-Benhamou et al. (1995) used intramuscular electrodes and determined that splenius capitis was active in extension and ipsilateral extension and inactive in flexion and contralateral flexion and rotation. As a result of these findings, the appropriateness of the use of surface electrodes to detect the activity of splenius capitis was questioned. Hence, surface electrodes are effective in recording muscle activity of large, superficial muscles however, crosstalk from adjacent muscles (e.g. sternocleidomastoid when trying to record splenius capitis) may question the validity of surface electrode measurements on small or deep muscles (Mayoux-Benhamou et al., 1995). The anatomical arrangement of muscles is responsible for the phenomenon of crosstalk and is unavoidable (Kramer et al., 2003).

McGill et al. (1996) used surface electrodes in an attempt to represent the muscle activation of the deep muscles of the lumbar spine. Psoas, quadratus lumborum, external oblique, internal oblique and transverse abdominis were represented by a single electrode placement and using RMS difference (difference between surface and intramuscular channels) and the coefficient of determination (R^2) it was concluded that a representation of deep muscles was possible with well-selected surface electrode locations as opposed to intramuscular electrodes. The authors concluded that their method favoured studies that

wish to avoid invasive procedures and which were willing to accept errors in the magnitude of 10-15% of MVC.

Stokes, Henry and Single (2003) investigated whether surface electrodes placed over multifidus could accurately represent muscle activity recorded from intramuscular electrodes inserted in multifidus. Linear regression (R^2) was used as the statistical method to determine the degree of correlation between the electrode sites ($R^2 = 0.64$). The authors determined that crosstalk was responsible for the observed degree of correlation as the surface electrodes were more sensitive to the adjacent longissimus rather than the underlying multifidus. The authors concluded that intramuscular electrodes were required for accurate measurement of multifidus muscle activity.

2.2.3 Description of Intramuscular Electrodes

Due to the invasiveness of intramuscular electrodes (fine-wire and needle) there has been a reluctance to use them in routine investigation. However, even if a muscle has a small superficial area (e.g. splenius capitis) intramuscular electrodes can be used to gather sensitive information (Mayoux-Benhamou et al., 1995). Ertekin, Celebisoy and Uludağ (2001) used needle electrodes with a 0.46mm diameter electrodes and a recording area of 0.07mm^2 to measure intramuscular activity of deep cervical muscles.

Different types of fine-wire electrodes have been used to record intramuscular EMG activity. Examples include 76μm platinum-iridium alloy wire with Teflon insulation (Wittek et al., 2001), woven platinum wire electrode pairs (Magnusson et al., 1999) and 0.15mm diameter copper bipolar wire electrodes (Mayoux-Benhamou et al., 1995). The fine-wire electrodes can be prepared in different ways such as stripping off the covering insulation. For example, the insulation may be cut 10mm from the tip (Wittek et al., 2001) or by passing two strands through a needle and removing the distal end of the tip, then burning and cutting the resulting wire to form 1-2mm of barb wire. This may then be utilised as a hook, and can effectively act as an anchor when the electrode is injected into the desired muscle (Mayoux-Benhamou et al., 1995). Fire-wire electrodes are usually

inserted via a hypodermic needle (Keshner et al., 1989; Magnusson et al., 1999; Mayoux-Benhamou et al., 1995; Wittek et al., 2001) and are typically implanted by a neurologist after sterilisation and local anaesthesia of the insertion areas (Magnusson et al., 1999). Nowadays, pre-packaged, sterile, bipolar electrodes are commercially available which means many of these procedures do not need to be performed.

2.2.4 Surface and Intramuscular Electrode Positioning in the Cervical Region

2.2.4.1 Trapezius

Trapezius functions as a prime mover during elevation and rotation of the scapula (McLean, Chislett, Keith, Murphy & Walton, 2003) and may have a role in head and neck extension and lateral bending (Keshner et al., 1989). EMG activity of trapezius can be recorded with different surface electrodes with studies including 4mm diameter Ag-AgCl electrodes spaced 10mm apart (Keshner et al., 1989), 6mm diameter Ag-AgCl electrodes (Choi & Vanderby, 1999) and square tin electrodes with a 4mm diameter placed longitudinally and spaced 5mm apart (Mayoux-Benhamou et al., 1995). Choi and Vanderby (1999) placed electrodes at the C4/5 level and similarly, Mayoux-Benhamou and co-workers (1995) placed surface electrodes over the belly of trapezius at the C4 level. Keshner and associates (1989) had subjects elevate their shoulders against a manual resistance to palpate trapezius and placed electrodes over the palpated muscle belly around the level of C6/7 and posterior to the insertion on the lateral third of the clavicle. Additionally in the above-mentioned study a bipolar needle electrode for trapezius was used with the authors inserting the electrode close to the midline (within 2cm) at the C1/2 level.

2.2.4.2 Levator Scapulae

Levator scapula elevates the scapula and its origin has been determined by Kamibayashi and Richmond (1998) via cadaver dissection. Levator scapulae originates at the transverse processes of C1 to C4, with fascicles that insert on the superior border of the

medial scapula, running inferolateral in parallel at angles between 30° and 45° from the midline. Previously, Quessier, Blüthner, Bräuer and Seidel (1994) and Schüldt and Harms-Ringdahl (1988) had placed surface electrodes between the posterior margin of sternocleidomastoid and the anterior margin of trapezius, pars descendens. Fine-wire electrodes have also been used to record the activity of levator scapulae, for example Magnusson et al. (1999) implanted the guide needle for a wire electrode at the C4 level, 4cms from the midline at an incline of 20 degrees towards the sagittal plane.

2.2.4.3 Splenius Capitis

Splenius capitis is thought to contribute toward extension, ipsilateral rotation and bending of the neck (Sommerich et al., 2000). Keshner et al. (1989) and Quessier et al. (1994) inserted intramuscular electrodes to investigate splenius capitis at approximately the C4 level. The exact location for electrode insertion was achieved by muscle palpation during restive head extension and lateral rotation (in the same direction of the muscle) after measuring 6-8cm lateral and 6cm superior to the bony protuberance at the level of C7.

Mayoux-Benhamou et al. (1995) horizontally inserted bipolar wire electrodes into splenius capitis 2cm from the midline at the C4 level with penetration depth of 1.45-2.6cm (median = 2.26cm) which was calculated using computer tomography (CT) to provide a cross-sectional image at the C4 level. Similarly, Magnusson et al. (1999) used fine-wire electrodes which were inserted into splenius capitis 26mm lateral to C3 at a depth of 27mm and the positioning of electrodes was confirmed on cadaver studies, which were dissected to confirm accuracy and sensitively of the insertion method.

2.2.4.4 Semispinalis Capitis

Semispinalis capitis is a primary head/neck extensor (Sommerich et al., 2000). Kramer et al. (2003) concluded that a high degree of reliability was present when puncturing the semispinalis capitis muscle when using regression equations to calculate the puncture angle and puncture depth. From using the hyoid bone as an anatomical landmark (due to

locating it with ease by palpation) for the top electrode, two electrodes were inserted in semispinalis capitis with the second electrode 0.5 cm below the first. The subject's neck circumference was used as a variable in the regression equation as the circumference is affected by muscle mass (which can be determined using computer tomography scans). Linear regression equations were then determined for fine-wire electrode positioning of semispinalis capitis, reporting the margin of error as 1.2 cm (puncture depth's median safe range) and 12.8° (puncture angle's median safe range).

The regression equations were: -

Predicted puncture depth $_{SSC}$ = circumference * 0.178 – 3.65
Predicted puncture angle $_{SSC}$ = circumference * 0.8 + 11.65

Keshner et al. (1989) and Quessier et al. (1994) placed intramuscular electrodes 2cm lateral to the midline at C1/2 level after the muscle belly was identified by the subject resisting head extension. The reasoning for semispinalis capitis' electrode positioning was that it was not covered by trapezius or splenius capitis from its insertion down to its C2 level. Magnusson et al. (1999) inserted fine-wire electrodes into semispinalis capitis 21mm lateral to C3 at a depth of 34mm, using calculations based upon MRI scans with accuracy confirmed by cadaver dissection studies.

2.2.4.5 Examination of the Deep Musculature in the Anterior and Anterolateral Regions

Morphological measurements (using CT/MRI techniques) of the deep cervical flexor muscles have been completed for application in computer modelling (Vasavada et al., 1998). Few attempts have been made using intramuscular EMG to measure muscle activation of the deep anterior and anterolateral muscles (Falla, Jull, Dall'Alba, Rainoldi & Merlett, 2003). The inaccessibility of the deep cervical flexor muscles is the main limitation preventing study of the anterior and anterolateral region by intramuscular EMG. Furthermore, the location of complex structures such as the trachea, carotid artery,

vagus nerve and lymphatics in close proximity to the intramuscular electrode placement sites, makes direct measurement highly complex and problematic (Falla et al., 2003).

As the deep cervical flexor muscles lie directly posterior to the oropharyngeal wall, Falla et al. (2003) developed and implemented a technique that utilised bipolar, silver wire electrode contacts attached to a suction catheter to measure the activity of longus colli and longus capitis without the need for intramuscular techniques. The device was inserted through the nose and placed on the posterior oropharyngeal wall and recorded activity via the mucosal wall. To minimise crosstalk the electrode was positioned at the C2/3 level where longus colli has its greatest cross-sectional area (Falla et al., 2003).

2.2.4.6 Determining the Correct Positioning of Intramuscular Electrodes

One of the difficulties with using intramuscular electrodes is determining the correct positioning in a target muscle. The correct positioning of fine-wire electrodes has previously been confirmed by cadaver studies, where specimens were dissected to confirm the accuracy and sensitivity of the insertion method (Magnusson et al., 1999). Further, real-time ultrasound hardware can be used by a skilled technician to visualise the extremely small wires however, it should be noted that ultrasound may compromise the sterility of the injected area (Kramer et al., 2003). A risk of intramuscular electrode insertion is that a haematoma may arise from a vascular lesion as the needle passes closely through deep, cervical arteries and veins. Insertion of intramuscular electrodes into semispinalis capitis is relatively safe as the needle passes only through one muscle, the trapezius muscle, thus lessening the chance of haematoma formation (Kramer et al., 2003).

2.2.5 Data Acquisition and Analysis

Data in EMG studies involving the neck are typically sampled at 1000Hz and are normally amplified and filtered (Brault et al., 2000; Magnusson et al., 1999, Wittek et al., 2001). Raw EMG data can be manipulated using different numerical techniques to

produce the desired method for assessing results. Dressler (2000) rectified EMG signals and divided the data into four second segments. The mean amplitude was measured for each segment and used to calculate the overall mean amplitude. Mayoux-Benhamou et al. (1995) manipulated their EMG data into grades of muscle activation. During free head movement raw EMG signals were scored into four grades by rating EMG activity, whilst during isometric contractions the amplified potentials were rectified, amplified and scored into four grades based on percentage of maximum signal intensity. Magnusson et al. (1999) sampled data in dynamic conditions and used a wavelet transform to eliminate unnecessary noise signal components as a signal is separated into different frequency components when using this technique. A wavelet transform is different from the common Fourier Transform as a signal is decomposed into both time and frequency domains (Magnusson et al., 1999). For functionality and application specific recording, Hamalainen and Vanharanta (1992) used a portable surface integrated EMG device, which measured in the range of 20-500Hz to record pilots performing aerial manoeuvres.

2.3 <u>The Use of Normalisation in Neck EMG Studies</u>

2.3.1 Normalisation Methods

The process of normalisation represents the raw EMG signal in a generalised way (Morris, Kemp, Lees & Frostick, 1998) and is done so that muscle activation can be compared between muscles and individuals. A maximum voluntary isometric contraction (MVIC) is commonly used to normalise electromyography (EMG) signals of the neck musculature (Choi & Vanderby, 1999; Kumar, Narayan & Amell, 2001; Moroney et al., 1988; Schüldt & Harms-Ringdahl, 1988). Studies of the deep muscles of the neck have been undertaken by various authors (Keshner et al., 1989; Magnusson et al., 1999; Wittek et al., 2001) however, to the author's knowledge, the reliability of normalisation procedures for intramuscular electrodes used for the neck region have yet to be examined. This is more than likely due to the invasive nature of the technique.

An alternative to MVIC normalisation is sub-maximal voluntary isometric contraction (sub-MVIC) normalisation. The reliability of sub-MVIC has been examined in the lumbar spine (e.g. Allison, Godfrey & Robinson, 1998; Dankaerts, O'Sullivan, Burnett, Straker & Danneels, 2004) and superficial anterior muscles (sternocleidomastoid and anterior scalenes) of the neck (Falla, Dall'Alba, Rainoldi, Merletti & Jull, 2002), but to the author's knowledge no such study has been undertaken pertaining to the posterior and posterolateral neck musculature. Sub-MVIC has been reported as more reliable in symptomatic subjects as they find difficulty in performing MVIC's (O'Sullivan, 2002). With respect to the neck it is not known whether any discrepancies in reliability are observed in intramuscular normalisation procedures between MVIC and sub-MVIC's.

Morris et al. (1998) investigated EMG in the shoulder muscles (supraspinatus, infraspinatus, subscapularis) using dual fine wire electrodes to determine the most appropriate method of normalisation. Pre-MVIC trials were used as well as normalising trial data to its mean or peak voltage in an effort to generate reproductive measurements of muscle activation. Trials normalised to peak voltage displayed the greatest reproducibility (4-13%), whilst MVC normalisation displayed extremely large variations (5-143%). An advantage of peak EMG is that it keeps EMG in the range of 0-100%, which is preferable when inputting data into certain neck models (Vasavada et al., 1998).

Mathiassen, Winkel and Hägg (1995) concluded that expressing amplitude normalisation as a percentage of a reference voluntary electrical activation (%RVE) or maximum voluntary electrical activation (%MVE) is a superior method. However, there are advantages and disadvantages for both methods (Sommerich et al., 2000). In an attempt to decrease the sensitivity of a contraction used for normalisation, Veiersted (1991) noted that an external load could be used however, the capacity to which the muscles are working would be unknown.

When a subject is unmotivated or unaccustomed to performing a MVIC, the validity of the measurement becomes questionable and this issue concerning normalisation has been expressed by Jensen, Vasseljen and Westgaard (1996). The authors stated that most

people were unaccustomed to making forceful neck extensions and therefore, one maximal contraction was not a reliable estimate, as practice whilst avoiding fatigue is needed to produce maximal efforts. However, repeated contractions may cause discomfort, potential injury and delayed muscle soreness (Veiersted, 1991).
There exists variability in MVIC data which may be attributed to the fact that the MVIC is difficult to achieve. Techniques that may improve a subject's ability to attain a MVIC include minimising a possible learning effect through the use of a familiarisation session, providing visual feedback and giving verbal encouragement (Ng, Parnianpour, Kippers & Richmond, 2003). These method have been reported to have an effect on the ability of a subject to achieve repeated, consistent, maximal values with low variability, for example consistent between-days torque values (ICC = 0.98) for the elicitation of MVC in axial rotation of the trunk (Ng et al., 2003).

2.3.2 Effect of Window Length on Normalisation Values

With respect to the neck, the effect of moving window length has previously been discussed in the literature (Jensen et al., 1996; Mathiassen et al., 1995; McLean et al., 2003) and it has been reported that the chosen length of the moving window elicits changes in amplitude of the average EMG signal. For example, McLean et al., (2003) investigated the effect of window length on final root mean square (RMS) values for activation of the upper trapezius muscle. Using different window lengths, no significant difference was found for window lengths between 200, 300, 400 and 500msec. A significant difference was found between window lengths of 100msec and 300, 400 and 500msec.

McLean et al (2003) concluded that window length had a significant effect on the calculated RMS amplitude however, window lengths smaller than 100msec were not investigated. Mathiassen and co-workers (1995) concluded that a 100msec moving window was the optimum choice for the processing of an EMG signal. However, the authors put forth no evidence for their conclusion and hence validity of their recommendations is needed to confirm the most reliable moving window. McLean et al.

(2003) and Mathiassen and co-workers (1995) concluded that window length should be standardised so research findings between studies could be compared. It should be noted the choice of moving window length that generates changes in EMG amplitude in the abovementioned methods were calculated using RMS and not a linear envelope.

Previous literature in the lumbar spine made use of a 25msec moving window (e.g. Dankaerts et al., 2004) however, the validity of, for example, a 25msec versus a 200msec window approach in the posterior and posterolateral neck muscles has yet to be examined. Considering the stochastic nature of the EMG signal, a 25msec window may affect reliability by inducing a considerably random deviation in the averaged amplitude. St-Amant, Rancourt and Clancy (1998) determined that the signal to noise ratio (SNR) increased with an increase in window length in a square root fashion (a higher SNR was more desirable) when window lengths of 2.45-500msecs were investigated in contractions made with the biceps and triceps muscles at differing intensities.

2.4 Summary

As neck pain in occupational tasks is a problem that involves a high cost to society, there is a need for the generation of a musculoskeletal model to aid in addressing this issue. Once a musculoskeletal model of the neck is completed, quantitative data on neck loading may be gained non-invasively for a wide range of occupational tasks. However, in order for the model to be valid, the deep muscles of the neck must be examined as they have potential to add to the moment generating capacity. This has been investigated in the lumbar spine but not the cervical spine. If EMG activity of the deep muscles can be predicted from the EMG of surface musculature at the posterior and posterolateral aspects of the neck, then collecting data of these deep muscles will be less complex as invasive methods will not be required.

To date, studies have been conducted to determine accurate positioning of electrodes for recording the activity of superficial and deep muscles in the cervical region. Many different methods of obtaining EMG data from deep muscles of the neck have been used

and fine-wire electrodes have been found to produce good results. At present, the use of surface electrodes positioned on superficial aspects of splenius capitis and semispinalis capitis is not recommended. The determination of reliable normalisation methods is also important if data is to be used to make future comparisons between for example, surface and intramuscular electrode sites. The reliability of MVIC and sub-MVIC normalisation methods, the effect of window length and the repeatability of intramuscular electrodes must be quantified before comparisons can be made. This will allow an accurate conclusion to be made as to whether EMG activity of the deep muscles can be predicted from the EMG of surface musculature at the posterior and posterolateral aspects of the neck.

CHAPTER THREE

3.0 MATERIALS AND METHODS

3.1 Subjects

Six male subjects with mean (±SD) age (31.3 ±7.5 yrs), height (179.3 ±4.3 cm) and mass (81.0 ±8.2 kg) were recruited as subjects for this study. Exclusion criteria were; current cervical injury, history of dizziness, fainting or symptomatic irregular heartbeat, exertion-induced aggravation of a severe headache, diagnosis, symptoms or risk of carotid or coronary artery disease, current diagnosis of high blood pressure or severe limitations in pulmonary function capacity (Sommerich et al., 2000). Ethical approval for the study was granted by the Edith Cowan University Human Research Ethics Committee and an informed consent document was signed by subjects prior to the commencement of data collection.

3.2 Experimental Protocol

Subjects performed three, five-second maximum voluntary isometric contractions (MVIC) and three, five-second 60%-submaximal isometric contractions (60%-MVIC) against a semi-rigid flat pad attached to the torque arm of a Cybex isokinetic dynamometer (Cybex 6000, Ronkonkoma, NY). Peak torque values rather than the EMG values from the MVIC trials were used as the basis for determining the 60%-MVIC trials. Therefore, a total of 48 contractions were performed with contractions made in extension and right lateral bending with the head in neutral (0°) and three non-neutral (20° flexion, 30° lateral bending and 35° extension) positions (see Table 2). As neck muscles have been shown to produce different forces/torques at different lengths (McLean et al., 2003) both neutral and mid-range head positions were examined. Due to the invasive nature of the study only unilateral recordings were measured and the right-side was chosen by default.

Head position was obtained by instructing subjects to rotate around the C4/5 joint in the desired direction. Head positions that occurred in the same plane as the intended contraction direction (e.g. extension with the head in 20° flexion) were measured with a goniometer once and set at that position using the isokinetic dynamometer for all trials. For all other head positions not in the same plane as the contraction direction (e.g. extension with the head in 30° lateral bend) the position of the head was measured before the commencement of each trial with a goniometer.

Table 2. Summary of contractions in experimental protocol.

		Direction		
Head Position	**Intensity**	Extension	Right Lateral Bending	
Neutral (0°)	MVIC	3	3	
	60%-MVIC	3	3	
Flexion (20°)	MVIC	3	3	
	60%-MVIC	3	3	
Lateral Bending (30°)	MVIC	3	3	
	60%-MVIC	3	3	
Extension (35°)	MVIC	3	3	
	60%-MVIC	3	3	
	Total	24	24	48

With respect to contraction-directions/head-positions, these were randomised to avoid an ordering affect. However, in each contraction-direction/head-position all three MVIC trials were performed before the three 60%-MVIC trials as the peak torque was needed to evaluate the 60%-MVIC. Physiological recovery was facilitated by allowing a two minute recovery between MVICs and a one minute recovery between 60%-MVICs.

3.3 <u>Data Collection</u>

Subjects were seated in a fabricated chair that utilised a five-point racing-car harness system so the neck was isolated as much as possible (see Figure 2). The semi-rigid flat pad was moulded slightly to the subject's head allowing resistance to be provided orthogonal to the intended movement in an effort to minimise extraneous non-planar movement and the axis of rotation of the torque arm was aligned to the subject's C4/5 level. The approximate centre of pressure of the pad for contractions made in right lateral bending was the most lateral point of the squamous portion of the temporal bone, directly superior to the external auditory canal. The centre of pressure of the pad for contractions in extension was the mid-point of the external occipital protuberance.

3.3.1 Dynamometry

During MVICs verbal encouragement was provided to each subject to ensure maximal effort. For 60%-MVIC efforts subjects were provided with visual feedback from a computer monitor that was positioned directly in the subject's line of sight for each contraction-direction/head-position to assist them in achieving the desired level of contraction. For each contraction-direction/head-position the peak torque over the three MVIC's was used to determine the 60%-MVIC torque for each sub-maximal contraction set. The resulting value was depicted as a line on a torque-history graph displayed by a customised software program.

<u>Figure 2:</u> Custom-manufactured, fabricated chair that features a five point racing-car harness system.

3.3.2 Surface and Intramuscular Electromyography

EMG activity was collected unilaterally (right side) from the posterior and posterolateral aspect of the neck utilising both surface and intramuscular electrodes. Intramuscular electrodes were inserted into semispinalis capitis (posterior) and splenius capitis (posterolateral) with surface electrodes positioned over trapezius (posterior) and levator scapulae (posterolateral).

Electrode positioning was chosen in an effort to combine aspects of previous studies involving mathematical neck modelling (e.g. Vasavada et al., 1998) and EMG techniques. Also, the electrode locations were chosen to record activity from over the muscles responsible for producing movement in the directions of extension and right lateral bending. With the head in the neutral position, the spinous processes of C4 and C5 were identified by palpation and the interspinous space marked. All intramuscular and surface EMG recordings were made from this level to ensure similar electrode placements for all subjects.

Intramuscular EMG recordings were made using bipolar fine-wire electrodes insulated with Teflon (50.8μm, Nicolet Healthcare, Madison, RI). The end of the wire was stripped of insulation which allowed isolated recording of EMG from the target muscle only. A small hook at the end of the fine-wire kept it in a stable position once inserted. Each subject's skin was sterilised and local anaesthetic (1% lignocaine) was injected subcutaneously.

Prior to intramuscular electrode insertion, the fine-wire electrodes were preloaded into a twenty-five gauge hypodermic needle to enable insertion (see Figure 3). Accurate anatomical localisation was achieved using ultrasound (Model SSA-220A - CAPASEE II, Toshiba Medical System, Japan) to visualise the local soft tissues. A 7.5MHz probe (PVG-720S) was used to optimise superficial soft-tissue resolution.

For semispinalis capitis the needle was inserted 2.0-3.0cm lateral to the midline, in the posterior-anterior direction. Under ultrasound guidance the needle was then advanced to within the fascial boundaries of the muscle, then withdrawn leaving the hook wires in place. For splenius capitis the needle was inserted 2.5-3.5cm lateral from the midline, aiming anteromedially and using ultrasound guidance as described above (see Figure 4).

Figure 3: Withdrawal of the hypo-dermic needle, leaving the fine wire electrodes inserted intramuscularly.

After the needle was removed the fine-wires were difficult to visualise on ultrasound due to their fine diameter. However, anatomical localisation was confirmed at the end of the test protocol using gentle traction on the fine-wires and visualising the movement of hyperechoic structures at the hook site. This was not done prior to testing due to the risk of straightening the hooks.

If there was evidence that the wires had migrated at the end of the test protocol then the trial was considered void. Using this method of exclusion, no trials were considered void in this study.

The fine-wire electrodes were taped to the skin at the puncture site. The non-insulated tips were attached to micro-grabbers (Nicolet Healthcare, Madison, RI) and the 1.25m lead was finished with a DIN-42-402 connector. This allowed direct compatibility to the sixteen channel Grass Amplifier Rack's electrode board (Nicolet Biomedical, Madison, RI) which was used to collect the raw signals. The micro-grabbers were also taped to the skin to inhibit the potential displacement of the fine-wires.

Following intramuscular electrode injection the subject's skin was thoroughly prepared by shaving, lightly abrading and cleaning with alcohol allowing 12mm diameter Ag/AgCl disposable surface electrodes (Uni-Patch, Wasbasha, MN) to be placed over trapezius and

levator scapulae. Electrodes for trapezius were positioned 1cm from the spinous process at the C4/5 level in a bipolar configuration and placed between the anterior border of trapezius, in line with muscle fibres. Electrodes for levator scapulae were determined by palpating midway between the anterior border of trapezius and the posterior border of sternocleidomastoid at the C4/5 level and placed in a bipolar configuration.

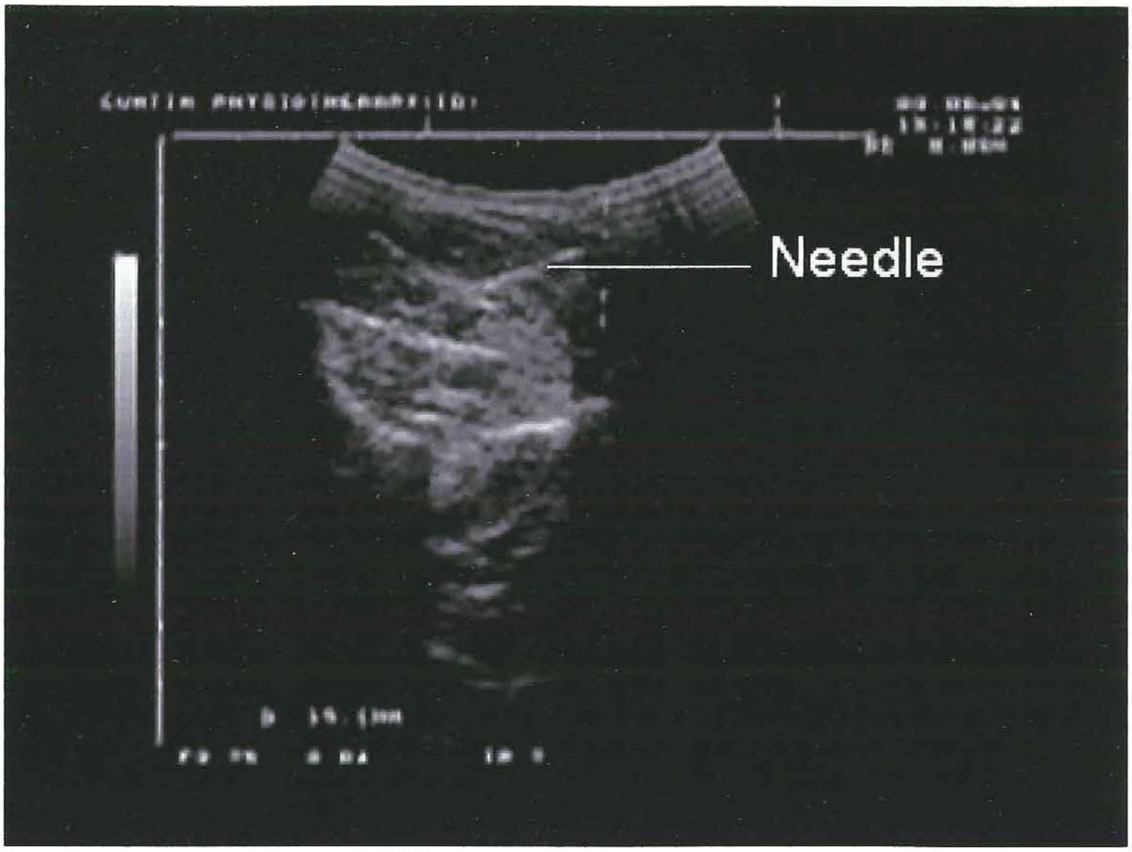

Figure 4: Posterolateral (splenius capitis) intramuscular electrode injection. Example of the needle in the neck on ultrasound.

Levator scapulae is situated superficially at the location of the posterior-triangle. Cadaver investigation prior to testing determined that positioning surface electrodes in line with the muscle fibres is prevented by the anatomical arrangement of the muscles at the posterior-triangle which overlay levator scapulae (namely trapezius, sternocleidomastoid and the scalenes). The inter-electrode distance of 20mm was chosen for both sites, as

recommended to avoid unstable recordings by not exceeding ¼ of the fibre length of the muscle (Freriks, Hermens, Disselhorst-Klug & Rau, 1999). Both surface and intramuscular channels shared the same ground placement reference of the clavicle, with the electrode placed medially towards the sternum to avoid contact with restraint straps of the fabricated chair.

All EMG and torque signals were sampled at 1000Hz with EMG signals also band-pass filtered at 10-1000Hz (Merletti, Farina, Hermens, Freriks & Harlaar, 1999). Data was recorded using a customised software program written in LabVIEW V6.1 (National Instruments, Austin, TX).

3.4 Reliability of EMG Normalisation Procedures in Posterior and Posterolateral Neck Muscles[1]

3.4.1 Introduction

Due to the lack of relevant research in the area it was necessary to perform a reliability study examining the best methods to normalise deep and surface muscle activation in the posterior and posterolateral neck muscles prior to data analysis. This study was conducted using the data with the head in the neutral position for extension and lateral bending contractions only (see Table 2).

3.4.2 Methods of Data Analysis for EMG Normalisation

3.4.2.1 Torque

As torque was collected at 1000Hz when typically it is measured at a lower sampling rate (e.g. 50-200Hz), noise due to over-sampling had to be removed. Raw torque data was

[*] This reliability study is currently in preparation for submission to the *Journal of Electromyography and Kinesiology*.

therefore processed using a customised Microsoft Excel macro where signals were low pass filtered at 4Hz using a fourth-order dual pass Butterworth digital filter.

To calculate the 60%-MVIC value for each contraction-direction/head-position, 60% of the peak torque value calculated from the second and third MVIC trial was used (see Section 3.4.4.1). This peak torque value was determined from a 1000msec window where the sum of squares (SS) (calculated from firstly, the 60% torque value pre-determined from the maximal torque trials and secondly, the actual torque output from the sub-maximal trial) was minimised.

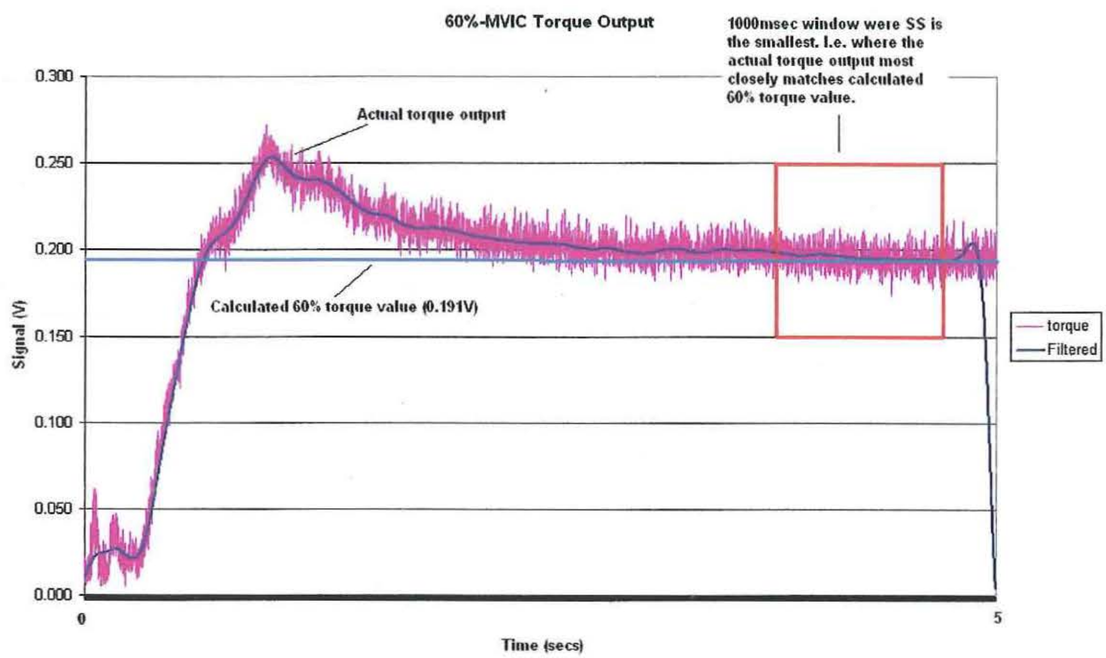

Figure 5: Example method of how 1000msec window was determined for sub-maximal torque. Firstly, the calculated 60% torque value was determined from the peak torque of the MVIC trials (e.g. 100% torque = 0.318V then 60% torque = 0.191V). A 1000msec moving window was then passed over the data to determine where the actual sub-maximal torque output matched the calculated 60% torque value the closest (using least sum of squares, SS). The light blue line is the calculated 60% torque value, the pink signal is the actual raw torque output containing noise due to over-sampling and the dark blue signal is the actual filtered torque output.

The location of this window within in each five second sub-maximal trial was recorded and later used as the area in which the average muscle activation for the sub-maximal trials was to be calculated. This was performed to provide uniformity to the data analysis and remove the sections of data where the subject was not actually producing 60% of maximal torque to increase the sensitivity of results (see Figure 5).

3.4.2.2 Electromyography

Raw data was processed using a customised Microsoft Excel program. EMG signals were demeaned, full-wave rectified and low pass filtered at 4Hz using a fourth-order dual pass Butterworth digital filter, to form a linear envelope.

To calculate peak EMG activation for the MVIC contractions in each contraction-direction/head-position, each channel was initially calculated by determining the peak value over 25, 50, 100, 150 and 200msec moving windows over the course of the entire trial. From this analysis each of these moving windows proved to be equally reliable (see Section 3.4.4.2) therefore, the 200msec window was subsequently used to calculate peak activation for this analysis due to the stochastic nature of the EMG signal.

For the 60%-MVIC trials, EMG activation was determined as an average from within the 1000msec window that was defined as the time over which the subject sustained 60% of MVIC (calculated by least SS for torque). Unlike the torque data, the EMG window was also padded (±200msec) to form a 1400msec window in an attempt to consider electromechanical delay (Nigg & Herzog, 1999) (see Figure 6). Once this window had been identified, maximum muscle activation for each EMG channel was also calculated using a 200msec moving window.

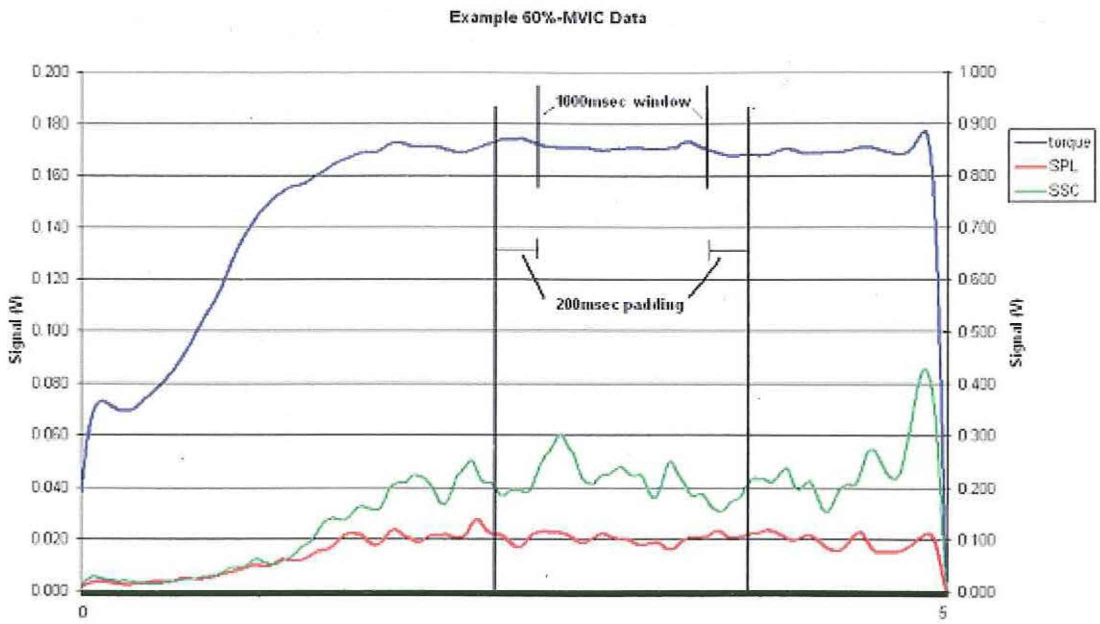

Figure 6: Example method outlining torque and EMG window determination for 60%-MVIC trials. Least sum of squares (SS = 0.110) was used to determine 1000msec window for sub-maximal torque (calculated 60%-MVIC torque = 0.174V). This was used to determine a 1400msec (1000msec and 200msec padding) EMG window (displayed on the primary axis). Filtered muscle activation of splenius capitis (SPL) and semispinalis capitis (SSC) is also shown on the secondary axis.

3.4.3 Statistical Analysis of EMG Normalisation Methods

Reliability of MVIC's and 60%-MVIC's in the posterior and posterolateral region of the neck was calculated to determine the within-subject variation using three indices of reliability; the intra class correlation co-coefficient (ICC), the relative standard error of measurement (%SEM) and the coefficient of variation (%CV). ICC values were calculated using the Statistical Package for Social Sciences Version 11.5 (SPSS V11.5) software and values were defined as 0.90-0.99, high reliability; 0.80-0.89, good reliability; 0.70-0.79, fair reliability; and 0.69 and below as poor reliability (Chiu & Lo, 2002).

ICC values were then entered into a Microsoft Excel spreadsheet, allowing SEM values to be calculated as follows: -

$$SEM = S_X \sqrt{1 - ICC}$$

(1)

Where, S_X is the pooled standard deviation of the activation in the respective contraction-direction/head-position trials. SEM values were used to calculate %SEM values as follows: -

$$\%SEM = \frac{SEM}{\overline{X}_i} * 100$$

(2)

Where $\overline{X}_i$ was the pooled mean of the activation in the respective contraction-direction/head-position trial.

Another method of analysing data for reliability used was the %CV. This method assesses variation between trials for one subject and contraction-direction/head-position at a time, and hence fits the characteristics for determining reliability well as it does not take into account pooled means and pooled standard deviations. The %CV is calculated as follows: -

$$\%CV = 100\% \cdot \frac{\sqrt{0.5 \cdot d^2}}{x_{pair}}$$

(3)

Where, d^2 is the squared difference between trial 1 and 2 and x_{pair} is the mean of the two measurements (Keller, Gunderson, Reikerås & Brox, 2003). In order to accommodate three trials as opposed to two, the %CV formula was manipulated for data with three trials. As d^2 was originally the squared difference between two trials, for three trials d^2

was calculated by taking the average of the squared differences between trials 1 and 2, trials 2 and 3 and trials 1 and 3. Also, x_{pair} became the mean of the three trials.

3.4.4 Results - Reliability of EMG Normalisation Methods

3.4.4.1 Reliability of Peak Torque Measurement in Extension and Right Lateral Bending

Although this section of the thesis is dedicated to examining the reliability of EMG normalisation procedures it is also important to determine the reliability of the torque measurements as they influence the EMG values due to the EMG-force relationship. ICC values of torque for MVIC trials were calculated for all three trials and also rejecting the first trial (i.e. for trials two and three). Rejecting the first trial (extension: ICC = 0.985, %CV = 4.8; right lateral bending: ICC = 0.949, %CV = 2.4) produced better ICC and %CV than accepting all three MVIC trials (extension: ICC = 0.982, %CV = 11.9; right lateral bending: ICC = 0.895, %CV = 3.7). The variability in the least the sum of squares value recorded for the 60%-MVIC torque histories trials were deemed acceptable (extension: SS = 0.192 and right lateral bending: SS = 0.079) (see Table 3).

Table 3. Evidence for rejecting the first MVIC trial based on torque data

	Extension			Right Lateral Bend		
	ICC	**%SEM**	**%CV**	**ICC**	**%SEM**	**%CV**
Trial 1,2,3	0.982	4	11.9	0.895	6	3.7
Trial 2,3	0.985	4	4.8	0.949	2	2.4

3.4.4.2 Reliability of Surface and Intramuscular EMG

To determine the repeatability of surface and intramuscular electrodes, both MVIC and sub-MVIC trials were investigated by comparing the peak EMG activation. As 25, 50, 100, 150 and 200msec window lengths displayed little deviation in means (±SD) across window lengths in both MVIC (extension: ICC = 0.937 ±0.000; right lateral bending:

ICC = 0.934 ±0.004) and sub-MVIC (extension: ICC = 0.956 ±0.000; right lateral bending: ICC = 0.915 ±0.000) trials, a 200msec moving window was used.

Table 4. Repeatability of surface and intramuscular electrodes

		Extension			Right Lateral Bend		
		ICC	**%SEM**	**%CV**	**ICC**	**%SEM**	**%CV**
MVIC	Surface	0.979	9	12.3	0.994	6	6.0
	Intramuscular	0.987	8	12.9	0.833	32	25.3
	Average	**0.983**	**8.5**	**12.6**	**0.914**	**19**	**15.7**
60%-MVIC	Surface	0.957	14	18.4	0.993	6	9.0
	Intramuscular	0.922	20	12.5	0.981	13	14.6
	Average	**0.940**	**17**	**15.5**	**0.987**	**9.5**	**11.8**

All muscles had highly reliable ICC values associated with them for both surface (MVIC: ICC = 0.986, %CV = 9.2; 60%-MVIC: ICC = 0.975, %CV = 13.7) and intramuscular (MVIC: ICC = 0.910, %CV = 19.1; 60%-MVIC: ICC = 0.952, %CV = 13.5) electrode positions (see Table 4).

3.4.5 Discussion - Reliability of EMG Normalisation Methods

In order to improve the reliability of generating a MVIC or sub-MVIC, methods which produce variations in torque output must be controlled. This can be achieved by adhering to standardised positioning of subjects and by immobilising the trunk when measuring from the cervical musculature. This prevents trunk and leg musculature assisting in generating torque. In the literature, good reliability of chosen testing apparatus has been reported through the use of adequate restraining devices. This was provided through the utilisation of two straps to fasten the trunk with subjects seated in a fixed sitting position (Ylinen, Rezasoltani, Julin, Virtapohja & Mälkiä, 1999) and the use of a four-point restraint system (Kumar et al., 2001). This study used a five-point racing-car harness attached to a customised fabricated chair as an effective means of immobilising the trunk, hence controlling for trunk contributions to torque.

The aim of using different window lengths to calculate the MVIC and 60%-MVIC values was to determine whether there is an effect on the reliability of the measurements. McLean et al. (2003) found that differing window lengths affected the magnitude of the calculated amplitude, leading to the conclusion that window length should be standardised in order to compare across studies. Examination of the 200msec window revealed no difference in reliability to that of using differing window lengths (25-150msec) and was chosen as the SNR increases with an increase in moving window length (St-Amant et al., 1998).

Relative SEM values were calculated but were deemed inappropriate in the determination of reliability for this study. %SEM may not be the ideal method to analyse error in this situation due to the nature of its calculation. The %SEM value is affected by changes in the pooled standard deviation and pooled average hence, a sample population with significant between-subject variation will result in a large %SEM. Therefore, a more sensitive approach would be to utilise a method to observe within-subject deviations, e.g. %CV.

3.4.5.1 Reliability of Peak Torque Measurements

Highly reliable within-day ICC values were recorded for peak torque in extension (ICC = 0.985) and right lateral bending (ICC = 0.949) when the first maximal calculation was disregarded as suggested by the literature (Jensen et al., 1996; Sommerich et al., 2000). In healthy controls, maximum torque values have been found to produce highly reliable ICC values in neck contractions (level of the axis of rotation was unspecified) of 0.98 in extension and 0.95 in right lateral bending (Chiu & Lo, 2002). Furthermore, force values recorded in a male population with the head in a neutral position (with rotation occurring around the C7/T1 level) have produced reliable average between-day ICC values of 0.951 and 0.905 for extension and lateral bending respectively (Kumar et al., 2001).

Jensen et al. (1996) stated that most people were unaccustomed to making forceful neck extensions therefore, one maximal contraction was not a reliable estimate, as practice whilst avoiding fatigue was needed to produce maximum values. The reasoning for rejecting the first trial is that if a subject is unaccustomed to performing a MVIC then the validity of the measurement becomes questionable (Mathiassen et al., 1995). However, it should be noted that repeated contractions may cause discomfort, potential injury and delayed muscle soreness (Veiersted, 1991) therefore, limiting the number of contractions.

3.4.5.2 Reliability of Surface and Intramuscular EMG Normalisation Methods

The repeatability of surface and intramuscular electrodes was found to be good among all trials when using ICC as the index of reliability. However, when using %CV to examine reliability, the between-trial variation was greater for intramuscular electrodes in right lateral bending for both MVIC and 60%-MVIC trials. This may suggest that the electrode position chosen to represent the intramuscular equivalent of the muscles responsible for right lateral bending is unstable, as in extension, %CV values for surface and intramuscular electrode positions exhibit less deviation. Reasons for the error seen in the intramuscular position for right lateral bending may be due to the possibility of not measuring the optimal signal if the electrode position is not close enough to the muscle fibres as fine-wire electrodes cannot be repositioned after insertion (Zennaro et al., 2002).

Netto and Burnett (2004) determined the reliability of EMG normalisation for 60%-MVIC trials without the use of visual feedback (instead they used verbal feedback) in extension (ICC = 0.83) and right lateral bending (ICC = 0.86) with the head positioned in neutral. Comparing these results to those obtained in this study it can be seen that providing visual feedback markedly increased the reliability of the procedure in both extension (ICC = 0.940, %CV = 15.5) and right lateral bending (ICC = 0.987, %CV = 11.8). This clearly outlines the importance of providing visual feedback to the subject. The average variation of the trials in this study was deemed to be 60 ±2.7% for extension and 60 ±2.2% for right lateral bending. This indicates that 60%-MVIC's are as reliable as MVIC's (extension: ICC = 0.983, %CV = 12.6; right lateral bending: ICC = 0.914, %CV

= 15.7) in the muscles of the posterior and posterolateral aspects of the neck in healthy controls.

3.4.6 Conclusions

The effect of changes in moving window length on the repeatability of EMG data collected from the posterior and posterolateral aspects of the neck revealed that a 25msec moving window elicited the approximately the same reliability to that of a 200msec window. It was decided to use the 200msec window in the deep versus surface electrodes study because of the stochastic nature of the EMG signal.

An isokinetic dynamometer (Cybex 6000) was used to reliably test subjects when the head is positioned in neutral and contractions are made in the direction of extension and right lateral bending. Also the use of visual feedback greatly increases the reliability of 60%-MVIC contractions. This study recommends the use of an additional monitor positioned directly in the line of sight of the subject for every sub-maximal contraction. Adhering to these recommendations increases the reliability of 60%-MVIC data, making it as reliable as using a MVIC for the purpose of EMG normalisation.

The reliability of surface and intramuscular electrodes was found to be good among all trials however, intramuscular electrodes were less reliable when compared to surface electrodes. Therefore, determination of the most reliable intramuscular electrode positioning for the muscles responsible for extension and right lateral bending of the neck may need to be investigated in future studies.

3.5 Data Analysis - Comparison of Deep and Surface Muscle Activation in Posterior and Posterolateral Neck Muscles

3.5.1 Methods of Data Analysis

3.5.1.1 Torque

Torque data was analysed as described in section 3.4.2.1. However, in this study all MVIC and 60%-MVIC trials in each of the contraction-directions/head-positions (neutral/non-neutral) were included for analysis.

3.5.1.2 Electromyography

EMG data was analysed as per Section 3.4.2.2. However, in this study all MVIC and 60%-MVIC trials in each of the contraction-directions/head-positions were included for analysis. From the results of Section 3.4.4, the first MVIC was rejected and the two subsequent MVIC trials were analysed to determine which trial contained the peak torque value. The EMG channels of the MVIC trial were then analysed using a 200msec window to determine the average peak activation. Once the average peak activation was determined for each EMG channel, it was used to normalise all data in both MVIC and 60%-MVIC trials.

3.6 Statistical Analysis

3.6.1 Methods of Determining Similarity Between Muscle Activation Patterns

Three methods were initially used to compare surface and intramuscular EMG activation patterns for the posterior and posterolateral neck muscles. Each set of signals were compared using the co-efficient of determination (R^2) (McGill et al., 1996; Stokes et al., 2003), the co-efficient of multiple correlation (CMC) (Kadaba et al., 1989; Mayagoitia,

Nene & Veltink, 2002) and the root mean of the squared differences (RMS difference) (McGill et al., 1996).

To compare the magnitude of surface and intramuscular EMG activity, the RMS difference was used (Mayagoitia et al., 2002, McGill et al., 1996). Intramuscular EMG results were used as a reference (Eq. (4)). Signals with similar amplitude will result in a low RMS. The formula for RMS was as follows: -

$$RMS = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(x_i(t) - x_j(t))^2}$$

(4)

Where $x_i(t)$ was the surface EMG activity at time point t and $x_j(t)$ is the corresponding intramuscular EMG activity at time point t. Further, the co-efficient of multiple correlation (CMC) was used to examine the similarity of the two EMG signals (Eq. (5)) (Kadaba et al. 1989; Mayagoitia et al. 2002).

$$CMC = \sqrt{1 - \frac{\sum_{j=1}^{N}\sum_{i=1}^{T}(Y_{jt} - \bar{Y}_t)^2 / T(N-1)}{\sum_{j=1}^{N}\sum_{i=1}^{T}(Y_{jt} - \bar{Y})^2 / (NT-1)}}$$

(5)

Where T was the number of data points and N was 2, the number of signals being compared (surface and deep). $\bar{Y}_t$ was the average of all data points for each signal (i) or (j) (Eq. (6)). Where (i) was the surface signal data point and (j) is the intramuscular signal data point being compared at a point in time.

$$\bar{Y}_t = \frac{1}{N}\sum_{j=1}^{N} Y_{jt}$$

(6)

And $\bar{Y}$ was the grand mean and is calculated by taking the average of all time points ((Eq. (7)).

$$\bar{Y} = \frac{1}{NT}\sum_{j=1}^{N}\sum_{t=1}^{T} Y_{jt} \tag{7}$$

As EMG amplitude was to be predicted at an instant in time, initial analysis of a set of data concluded that R^2 failed to determine absolute differences in signal magnitude and therefore was rejected as a means of signal comparison.

From further analysis of an initial set of data it was found that signals with significantly differing amplitudes at certain points in time led to the CMC equation becoming somewhat unstable. From viewing Equation (5) it can be observed that data that results in the numerator being greater than the denominator equates to the square root of a negative number, and hence no comparable data is gained through this method. However, with RMS difference no matter how widespread the data, information can be gained using this method.

As the aim of this study was to predict the difference in amplitude of two waveforms at a moment in time the RMS difference provided the best indication of predicability as a quantitative assessment of amplitude difference is obtained by using this methods. Therefore, it was decided that R^2 and CMC would be rejected in favour of RMS difference as the single, most effective means by which a statistical analysis would be performed on the data to determine the similarity between surface and deep EMG activity in the posterior and posterolateral aspect of the neck.

CHAPTER FOUR

4.0 RESULTS

4.1 Comparison of EMG Activation Between Deep and Surface Musculature in the Posterior and Posterolateral Aspects of the Neck

RMS difference was used to determine the closeness in amplitude of deep and surface muscle activation for the posterior and posterolateral neck muscles in neutral and non-neutral head positions. RMS values of zero represented signals that are identical in amplitude, whereas increasing values indicated increasing differences in amplitude. The mean (±SD) of the RMS values are shown in Tables 5 and 6 for the posterior and posterolateral neck muscles respectively. A trend observed was that the posterior muscles displayed a lower RMS value than the posterolateral muscles in MVIC and 60%-MVIC trials in both extension and right lateral bending.

Only descriptive data has been reported in the results section and this has been done for the following reasons. The research questions were to determine if; muscle activity from semispinalis capitis (a representative posterior deep muscle) can be predicted from muscle activity of trapezius (a representative posterior superficial muscle), and whether muscle activity from splenius capitis (a representative posterolateral deep muscle) can be predicted from muscle activity of levator scapulae capitis (a representative posterolateral superficial muscle). This could be examined in many conditions in this study namely; in neutral and non-neutral head positions and in maximal and sub-maximal isometric contractions. Rather than determine whether or not there were significant differences in RMS values in deep and superficial EMG activations between these conditions (neutral verus non-neutral head positions and maximal versus sub-maximal contractions) it was more important to determine the magnitude of these differences. In stating this however, it is difficult to determine what level of RMS difference between deep and superficial EMG activity is acceptable. This cannot be known until a sensitivity analysis using a musculoskeletal model is performed and a comparison of variables such as net torque and

compressive and shear forces is undertaken. Furthermore, the likelihood of finding any significant differences would be minimal due to the relatively large standard deviations and low subject numbers utilised in this study.

To examine the effect of EMG patterns and amplitude on RMS values, five random trials have been graphed (Figures 7-11). It can be observed that changes in the magnitude of the RMS values were consistent with changes in the closeness of amplitude between the deep and surface signals.

Table 5. Means and standard deviations for RMS differences (%MVIC) of activation between deep and surface muscles for the posterior neck muscles in extension and right lateral bending.

	Extension		**Right Lateral Bending**	
Head Position	**MVIC**	**60%-MVIC**	**MVIC**	**60%-MVIC**
Neutral	19.1 (7.4)	17.2 (7.7)	22.7 (6.9)	11.1 (2.6)
Flexion (20°)	16.2 (5.9)	15.3 (6.8)	24.3 (11.2)	24.2 (21.2)
Extension (35°)	27.5 (14.8)	21.0 (11.5)	24.1 (12.6)	25.0 (22.0)
Lateral Bend (30°)	19.0 (8.6)	15.2 (8.1)	23.3 (14.1)	11.9 (5.5)

Table 6. Means and standard deviations for RMS differences (%MVIC) of activation between deep and surface muscles for the posterolateral neck muscles in extension and right lateral bending.

	Extension		**Right Lateral Bending**	
Head Position	**MVIC**	**60%-MVIC**	**MVIC**	**60%-MVIC**
Neutral	27.0 (21.8)	22.2 (8.9)	24.9 (12.0)	17.7 (5.7)
Flexion (20°)	26.7 (18.8)	28.7 (16.4)	20.9 (9.4)	16.7 (8.4)
Extension (35°)	26.8 (15.8)	23.8 (10.3)	29.8 (23.0)	22.0 (6.6)
Lateral Bend (30°)	22.8 (9.1)	25.7 (10.0)	26.2 (11.4)	20.1 (7.9)

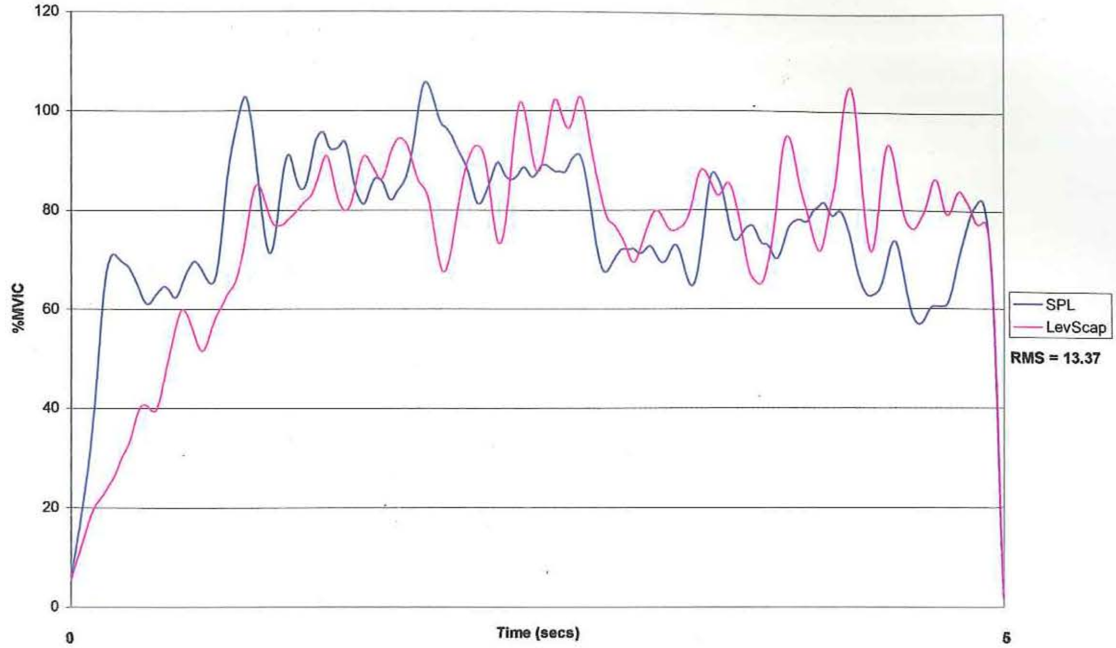

Figure 7: Comparison of the muscle activation patterns in a MVIC for the posterolateral muscles of the neck. Dark blue line is splenius capitis (SPL) and pink line is levator scapulae.

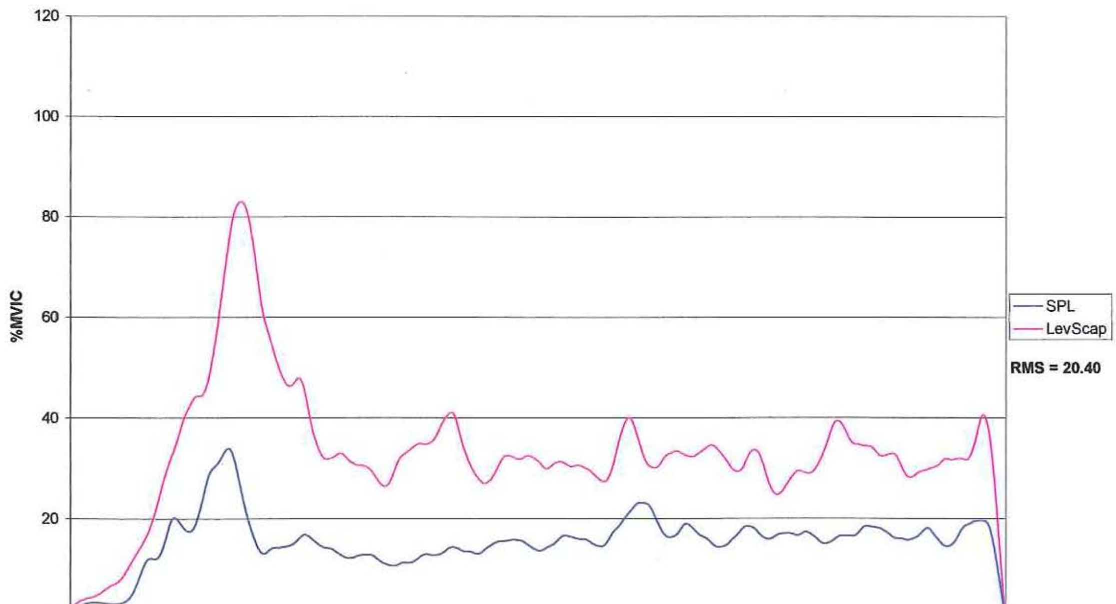

Figure 8: Comparison of the muscle activation patterns in a 60%-MVIC for the posterolateral muscles of the neck. Dark blue line is splenius capitis (SPL) and pink line is levator scapulae.

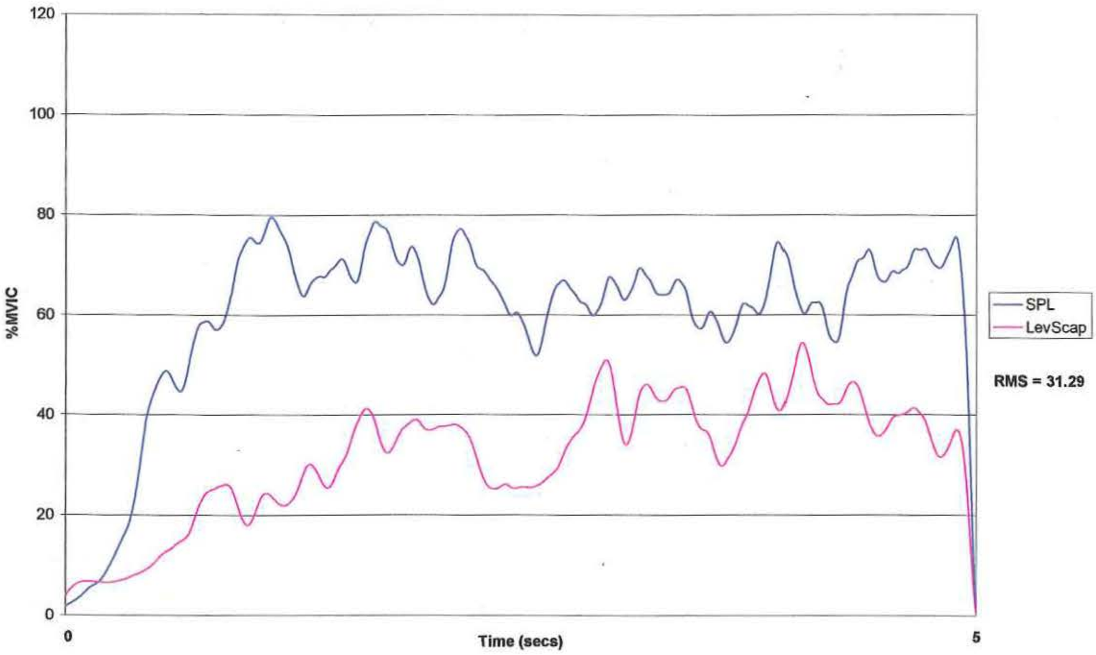

Figure 9: Comparison of the muscle activation patterns in a 60%-MVIC for the posterolateral muscles of the neck. Dark blue line is splenius capitis (SPL) and pink line is levator scapulae.

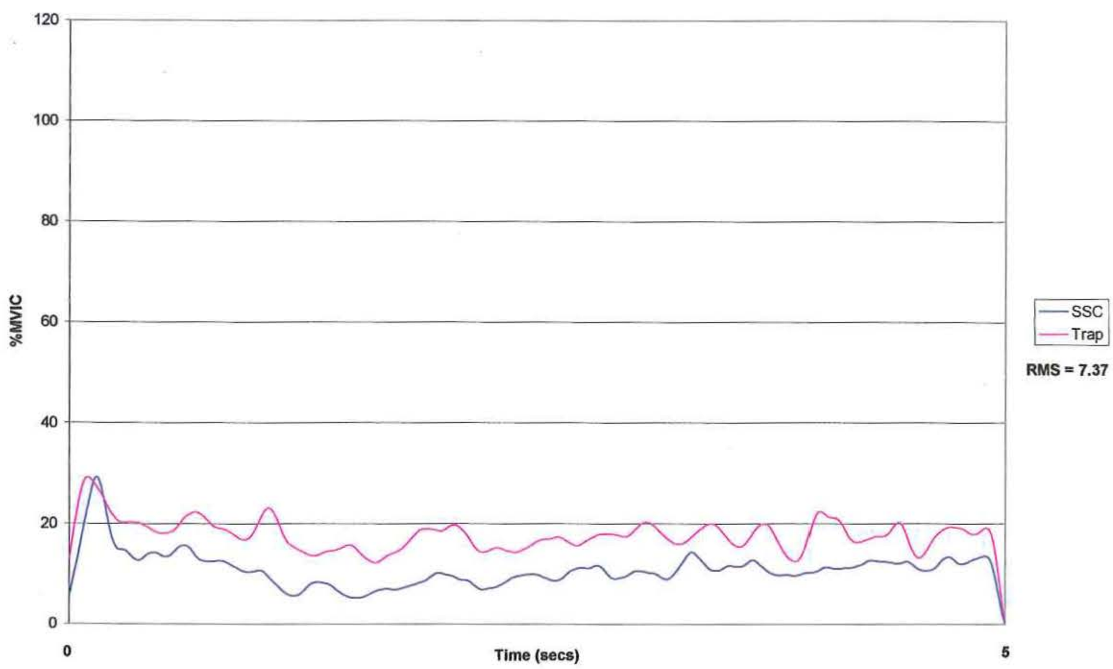

Figure 10: Comparison of the muscle activation patterns in a 60%-MVIC for the posterior muscles of the neck. Dark blue line is semispinalis capitis (SSC) and pink line is trapezius.

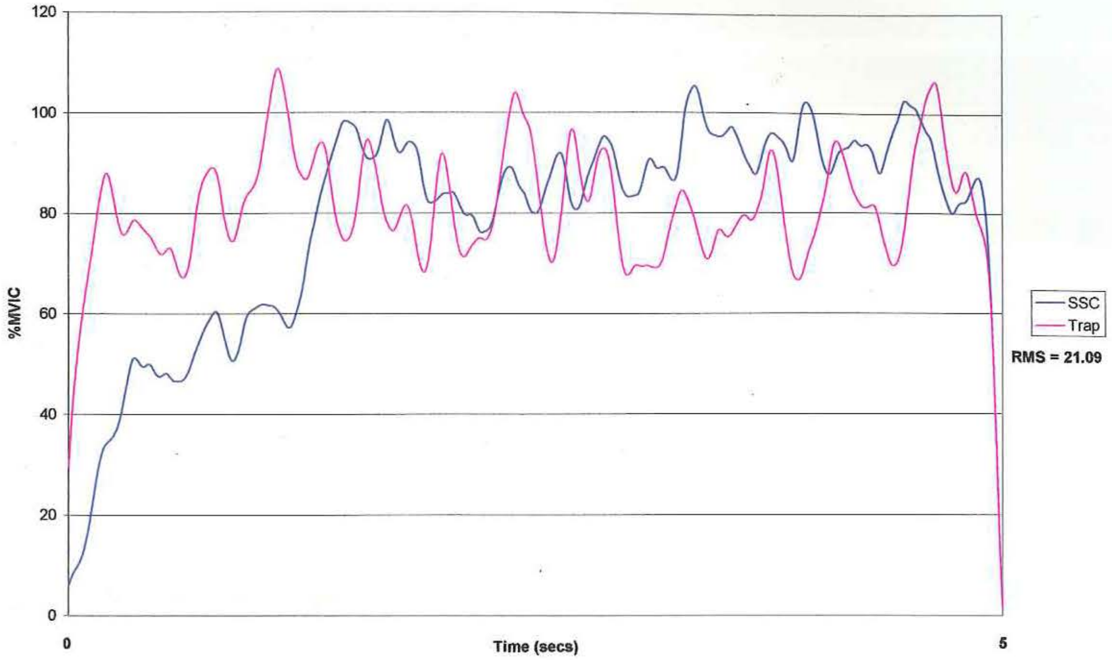

Figure 11: Comparison of the muscle activation patterns in a MVIC for the posterior muscles of the neck. Dark blue line is semispinalis capitis (SSC) and pink line is trapezius.

CHAPTER FIVE

5.0 DISCUSSION

5.1 Comparison of EMG Activation Between Deep and Surface Musculature in the Posterior and Posterolateral Aspects of the Neck

The hypotheses that muscle activation measured by surface electrodes cannot represent activation of the deep musculature in the posterior and posterolateral aspects of the neck can only be accepted or rejected once an appropriate level of RMS difference is determined. Using McGill and co-workers' (1996) study of the lumber spine as a guide, the degree of appropriate RMS difference is in the range of 10-15% however, only if one was willing to accept these differences. RMS differences in the magnitude of 10-15% may be acceptable in some modelling situations and also for clinical situations where a clinician may want to know if a muscle is activated or not (McGill et al., 1996). Given the invasive nature of the study, this level of difference may also be satisfactory as not accepting it requires the use of invasive procedures and the liabilities associated with these methods.

The mean of the RMS difference values between surface and deep musculature in this study was 19.8 %MVIC for the posterior aspect of the neck and 23.9 %MVIC for the posterolateral aspect of the neck. Given these values it is concluded that activation of the surface muscles does not represent activation of the deep musculature in the posterior and posterolateral aspects of the neck as in a majority of cases as the mean values are above the 10-15% cut-off as outlined in McGill and associates (1996) study (see Figure 12). If using the acceptance level of difference as the arbitrary range of 10-15%, then only two out of the thirty-two contraction-direction/head-position types are acceptable. Furthermore, it can be seen that the magnitude of RMS difference is contraction-direction, head-position and site dependent.

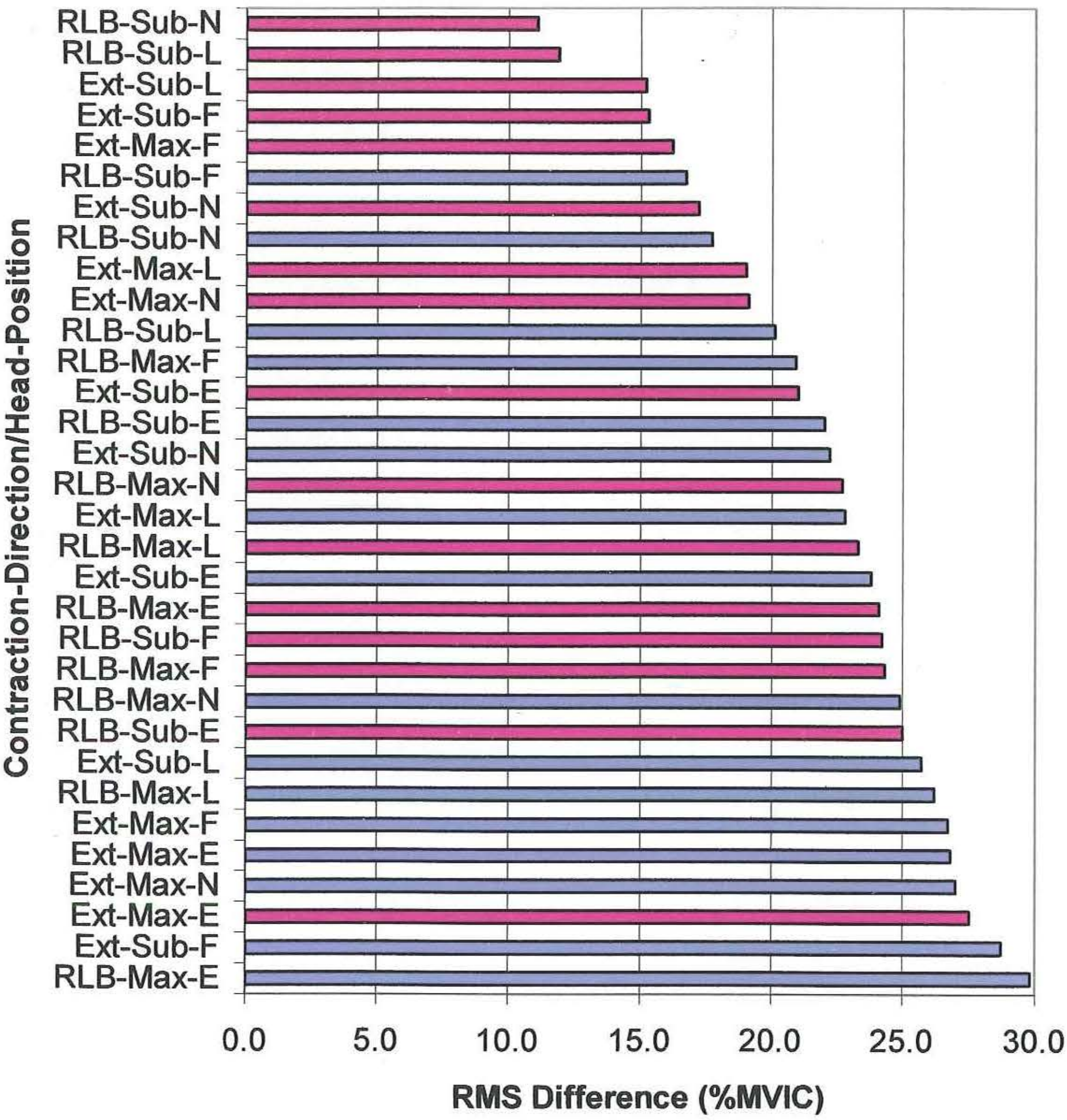

Figure 12: Mean RMS difference values of EMG activation between deep and surface musculature in the posterior and posterolateral aspects of the neck for all contraction-direction/head-position trials. Trial type is defined as follows; contraction direction (Ext = extension, RLB = right lateral bending), intensity (Max = maximal, Sub = sub-maximal) and head position (N = neutral, F = flexion, E = extension, L = lateral bending). Pink horizontal bars represent posterior neck muscle comparisons and blue horizontal bars represent are posterolateral neck muscle comparisons.

The implications of this conclusion are that invasive procedures would have to be used if an accurate representation of activity from the deep neck muscles is to be recorded. Although not encountered in this study, infection may arise from the use of intramuscular electrodes, ranging from the procedure of injection of fine-wires to the use of ultra-sound at the injection site which is used to visualise the extremely small wires. Another risk of intramuscular electrode insertion is haematoma formation that may arise from a vascular lesion as the needle passes closely through deep, cervical arteries and veins (Kramer et al., 2003).

Fine-wire electrode migration is a problem related to intramuscular procedures, which McGill and co-workers (1996) associated with their study. Intramuscular electrode migration was controlled for in this study by examining the position of electrodes at the completion of the experimental protocol. By using gentle traction on the intramuscular wires and visualising the movement of hyperechoic structures at the hook site, it was determined whether the electrodes had moved during testing. If there was evidence that the wires had migrated at the end of the test protocol then the trial was considered void. Using this method of exclusion, no trials were considered void in this study.

The implications associated with the use of intramuscular EMG electrodes highlights a limitation of this study that being; only the posterior and posterolateral muscles of the neck could be safely examined. Examination of the deep musculature in the anterior and anterolateral regions of the neck is problematic. This is due to the inaccessibility of the deep cervical muscles coupled with the location of complex structures, such as the trachea, carotid artery, vagus nerve and lymphatics in close proximity to the intramuscular electrode placement sites.

One method that has been used to overcome this problem was the creation of a suction catheter used to measure the activity of longus colli and longus capitis (Falla et al., 2003). Whilst surface electrodes were still used, this technique did not require the need for intramuscular techniques. As the anterior and anterolateral aspects of the neck were not examined in this study, the data collected pertains to only the posterior and posterolateral

aspects of the neck. Therefore, it cannot be determined whether activity of surface muscles represents activity of deep muscles in the anterior and anterolateral aspects of the neck.

The variation in the amplitude of muscle activity determined from the RMS difference may have been due to the way the surface and intramuscular electrodes function. As a surface electrode detects activity over a wide area, it may also detect the signal from the same motor units as the fine-wire electrodes and conversely fine-wire electrodes will not pick up activity of the entire muscle (Mayoux-Benhamou et al., 1995). Therefore, differences in EMG activity may be accounted for by surface electrodes recording activity representative of the whole muscle and the fine-wire electrodes recording activity specific to the location of the intramuscular electrode.

It can be observed that the RMS difference was lower in 60%-MVIC trials (average RMS = 19.8 %MVIC) compared to MVIC trials (average RMS = 23.8 %MVIC). A noticeable discrepancy was observed more in the initial portion of the contraction in MVIC trials than 60%-MVIC trials (see Figures 7-11). In comparing the deep and surface EMG signals it may have been beneficial to only compare the portion of the signal where the maximum activation of the signal occurred, if after all, that is where a value is obtained that represents the signal (e.g. peak activation from a 200msec window). For example, the RMS difference of Figure 11 was 21.09% however, computing the RMS difference between 1.5secs and 5.0secs (where the ramping has ceased) produced a RMS difference of 12.98%. This decrease in RMS may be significant in achieving a level of error that is desirable or acceptable. However, this approach does not consider the fact that the EMG activation patterns do not closely match in the ramping period of contraction. As all contractions in this study were isometric in nature, the RMS difference when concentric contractions are considered may give differing results. Specifically, if surface electrodes were used instead of intramuscular electrodes in dynamic contractions, there may be a greater magnitude of RMS difference.

5.2 Application to Musculoskeletal Modelling

In EMG-driven graphically based neck models (e.g. SIMM) (Vasavada et al., 1998) a number of muscles are represented through detailed anthropometric investigation (e.g. Kamibayashi and Richmond, 1998). Each task that is analysed in this model requires these muscles to be provided with an muscle activation profile. A number of these muscles are deep therefore, this provides the modeller with a problem as EMG activation profiles are difficult to attain without the invasive EMG procedures. Hence, the question raised in this investigation was to determine if EMG activation measured using surface electrodes could be used to represent EMG activity of deep musculature in the posterior and posterolateral aspects of the neck.

Even though it has been determined that the activation of surface musculature does not represent activation of the deep muscles at the posterior and posterolateral aspects of the neck, the effect of this RMS difference on variables output by this model can be observed through a sensitivity analysis as described below.

A forward simulation could be performed with the data generated from this study. In the current study, isometric contractions were generated against the torque arm of an isokinetic dynamometer therefore, the actual torque output is known and this could be used as a gold standard. Firstly, the muscle activation recorded by the surface EMG electrodes would be used to represent activation of the deep musculature. This data could be input into the neck model which would then output predicted torque data. The torque data that the model predicts, using activations of surface musculature to represent activations of deep musculature, could then be compared to the actual torque data produced in this study. Therefore, a functional error would be obtained by examining differences in torque data. Secondly, the measured EMG activation of the superficial muscles and the measured EMG activation of the deep muscles could be input into the model. The predicted torque values will be compared to the actual torque values produced in this study hence, this would provide a secondary validation of the neck model.

5.3 Conclusion

In order to determine whether EMG activation patterns from surface musculature can be used to represent EMG activity from deep muscles in the posterior and posterolateral aspects of the neck a criterion must be determined to ascertain a suitable RMS difference. Using McGill and co-workers' (1996) study of the lumbar spine as a guide, RMS differences above 10-15% MVIC are not acceptable therefore, it can be concluded from this study that EMG activity of the surface muscles does not represent activity of deep musculature in the posterior and posterolateral aspects of the neck.

5.4 Recommendations for Further Research

To further investigate the comparison of EMG activation between surface and deep muscle equivalents more research is needed. Questions arising from this investigation may include: -

i) The determination of the functional error level from predicted and actual torque output by performing a forward simulation.

ii) The determination of an appropriate method to compare EMG activity from surface and deep muscle equivalents in the anterior and anterolateral region of the neck. This may include implementing a suction-catheter type procedure (e.g. Falla et al., 2003).

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APPENDIX A

DOCUMENT OF INFORMED CONSENT

INFORMATION FORM

Deep Muscle Function in the Cervical Spine: Application to Musculoskeletal Modelling

Summary

This study will investigate the activation of the deep and surface neck musculature during static (isometric) contractions of the neck. The reason for this is that we are improving the validity of a musculoskeletal model of the head and neck for analysis of occupational tasks that have a high rate of injury and high cost to society.

In this study you will be asked to perform a total of 48 isometric contractions of your neck muscles. These contractions will be performed maximally (100%) and sub-maximally (60%) and will be done in four different head positions (neutral and non-neutral). Data from a total of four electromyography (EMG) electrodes will be collected in this study. These will measure the muscular activity when you perform an isometric contraction. Two of these electrodes will be surface electrodes and will be positioned near the back (trapezius) and side (levator scapulae) of your neck. The other two electrodes will be fine wire electrodes and these will record deep muscle activity. These electrodes are inserted into the deep muscles that correspond to the abovementioned surface muscles of the neck. These electrodes need to be inserted using a hypodermic needle after injection of a local anaesthetic. This will be performed by an appropriately trained medical doctor (neurologist) with the use of an ultrasound machine.

Risk and ethical considerations

You will need to be prepared for electromyography by shaving your neck and slight exfoliation of the skin. The surface electromyography poses no risk to the subject. The fine wire electrodes however are invasive and require an injection of local anaesthetic and a hypodermic needle as described above. Although the use of fine-wire (intramuscular) electrodes is a standard clinical procedure for diagnosis of neural problems, there are a number of risks involved. There is some discomfort as they are inserted. There is also a slight risk of bleeding and a possibility of delayed infection. There is also a small risk that a wire fragment may break off and remain in the muscle. This rarely happens however, if this does occur, there is a possibility that surgical removal may

be required. These various risks are minimal when the recordings are practiced by an experienced clinician. The electrodes are not being inserted near any nerves or major blood supply. The fine-wire electrodes will be inspected after removal to ensure they have not remained in the muscle. Ultrasound will also be used to ensure no portion of the electrode has been left behind after removal.

No direct comparisons between different individuals participating in the study will be made at any stage of the testing. Analysis of data will be made on a group basis with means and variances compared between another group. You are therefore not in competition with any other individuals in the study and will in no way be made to feel that your results are inadequate or incorrect.

All personal information and test results recorded will remain confidential and will not be used for any purpose other than the current study. Moreover, no data analysis will include your name or information that may identify you specifically as a subject. You will be free to withdraw from this study at any stage and for any reason without prejudice.

Requirements
Should you have any questions relating to any of the information provided above, please feel free to contact the undersigned for a further explanation. If you have any concerns about this research, or would just like to speak to an independent person, you may contact the Professor Robert Newton on telephone 9400 5711.

Yours Sincerely,

Dr Angus Burnett
School of Biomedical and Sports Science, Edith Cowan University
100 Joondalup Drive, Joondalup WA 6027
Phone: 9400 5860 E-mail: a.burnett@ecu.edu.au

DOCUMENT OF INFORMED CONSENT

Deep Muscle Function in the Cervical Spine: Application to Musculoskeletal Modelling

I ______________________________________ have read all of the information contained on this sheet, have completed a medical questionnaire and have had all questions relating to the study answered to my satisfaction.

I agree to participate in this study realising that I am free to withdraw at any time, for any reason without prejudice.

I agree that the research data obtained from this study may be published, provided I am not identifiable in any way.

Participant ______________________________ Date ____________________

Investigator ______________________________ Date ____________________