CERVICAL SPINE MOTION DURING PATIENT TRANSFER ONTO A LONG SPINE BOARD

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Abstract

The initial management stages of a suspected spinal cord injury are crucial. Currently there is a void in the literature with regards to the proper timing of neck realignment for prone patient transfer methods. The purpose of this study was to determine if the timing of neck realignment and/or the size of the victim will have an influence the amount of cervical spine motion during the prone log roll technique. A team of five Athletic Therapists performed 18 log rolls (9 on two trained "victims"), randomly correcting neck realignment AFTER and DURING the roll as well as the timing of their choice, for both supine and prone conditions. Motion of the cervical spine was collected using accelerometers and electromyography (EMG) was used to collect muscle activity of neck stabilizers. Comparisons were made for range, additional motion, theoretical minimal required motion and maximum EMG values. There were no significant differences found for the timing of neck realignment for motion and muscle activation, but there were significant differences found between male and female victims for both motion and muscle activation. These findings will help enhance knowledge of transfer techniques as well as help develop proper training techniques for primary stage management personnel.

Dedication

I would like to dedicate this thesis to my family for supporting me through all of my educational endeavours. I would especially like to thank my loving husband Matt and my son Jakob, who have inspired me and encouraged me to strive for greatness.

Acknowledgments

I would to start by thanking my advisor Dr. Janessa Drake for all of her guidance and enthusiasm throughout this whole process. I would also like to thank my mentor Dr. Frances Flint for helping me to realize my potential in the field of Athletic Therapy. Finally, I would like to especially thank Graham Mayberry for putting up with my constant demands and keeping a smile on his face the whole time I had the opportunity to work with him. It was an absolute pleasure.

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Abbreviations

CAT(C)	Canadian Certified Athletic Therapist
SCI	spinal cord injury
3D	three dimensional
EMG	electromyography
SCM	sternocleidomastoid
LSCM	left sternocleidomastoid
RSCM	right sternocleidomastoid
LES	left erector spinae
RES	right erector spinae
LUT	left upper trapezius
RUT	right upper trapezius
FH	forehead
MEMS	micro-electro-mechanical systems
sEMG	surface electromyography
nEMG	needle electromyography
MVC	maximum voluntary contraction
%MVC	percentage of maximum voluntary contraction
SD	standard deviation
TMRM	theoretical minimal required motion
f	female
m	male

1. Scope of Thesis

Catastrophic spinal cord injury (SCI) is not common, but up to 25% of secondary complications occur during the initial management stages (Conrad et al., 2012; Del Rossi et al., 2004).. It is defined as "a structural distortion of the cervical spinal column associated with actual or potential damage to the spinal cord" (Banerjee et al., 2004, p. 1007). SCI has devastating outcomes, including irreversible neurological deficits and premature morbidity, as well as staggering associated medical costs (more than $68 000/year for paraplegic individuals) (Dryden et al., 2004). Great care must be taken during the initial management stage of a suspected SCI because improper care can increase symptoms and worsen the injury (Banerjee et al., 2005). According to Cusick and Yoganandan (2002), although the initial injury occurs within milliseconds, any changes to alignment of the spine after the initial injury (i.e. during transfer onto a long spine board) may result in detrimental changes to the initial injury state of the spine. This makes SCI injury management a very important area of research (Blackham et al., 2009; Swartz et al, 2009). The initial management stage involves transferring the victim onto a long board while maintaining manual in-line stabilization of the cervical spine to eliminate potential exacerbation of the injury and secondary complications (Del Rossi et al., 2004). The definition and effectiveness of placing the spine in "neutral position" prior to immobilization remains unclear amongst researchers (Blackham et al., 2009; DeLorenzo et al., 1996), despite being arguably the most crucial stage in the transfer protocol.

Currently, there is excellent research describing the process of acute injury management before and after the transfer onto the board. There is a void however, regarding whether a neck that is not found in neutral alignment should be realigned during or after the transfer process for those found prone or supine. These studies do not provide an accurate representation of how a live subject would react to a suspected SCI and they also tend to focus on the whole spine, examining head movement relative to the torso, rather than on movement specifically at the neck (Conrad et al., 2012). The success of the transfer is dependent on the skill of those involved and there is a chance that the individual responsible for maintaining manual stabilization may not be able to ascertain contraindications for realignment (i.e., spasm, crepitus, and increased neurological symptoms) safely and effectively during the transfer process despite advanced training.

This study will use health care professionals, specifically Certified Athletic Therapists, performing a prone patient transfer technique on live, healthy models to quantify the amount of cervical spine movement, specifically axial rotation and lateral translation that occurs at the neck during this transfer. The target audience for this study will be those professions who primarily deal with sports related injuries.

1.1 Research Questions

The purpose of this thesis project was to quantify and assess the amount of angular motion and muscle activation in the cervical region during a prone log roll transfer onto a spine board.

The following questions will be addressed through this study:

1. Does the timing of neck realignment, either DURING or AFTER positioning on a long spine board, modify the amount of cervical spine motion and muscle activation during prone boarding?
2. Do the weight and size of the patient play a role in the amount of cervical spine motion?

1.2 Hypotheses

This study will capture three-dimensional (3D) motion occurring at the cervical spine during a prone log roll relative to the trunk during movement onto a spine board as well as muscle activity in three major neck stabilizers.

With these measures, the following hypotheses will be tested:

1. The timing of neck realignment does modify the amount of cervical spine motion and muscle activation during boarding. Specifically, alignment of the neck DURING the transfer will cause less motion and muscle activation as those stabilizing the head will not be able to maintain stabilization in the position found for the duration of the roll.
2. The weight and size of the patient does not play a role in the amount of cervical spine motion during transfer onto a spine board.

2. Review of Literature

The following section reviews relevant literature providing necessary background information to help set a solid knowledge foundation regarding cervical spine injury. More specifically, anatomy of the cervical spine, mechanisms of cervical spine injury, classification of cervical spine injury and current methods in patient transfer will be discussed. An outline of general methodology will be introduced including information on the use of accelerometers and electromyography (EMG). Methodology specific to this study will be discussed further in Section 3 (p.32).

2.1 Anatomy

2.1.1 Vertebral Column

The spinal column consists of 26 vertebrae and is divided into five regions: cervical (containing seven vertebrae), thoracic (containing 12 vertebrae) and lumbar (containing five vertebrae) and also includes the sacrum and the coccyx (McKinley & O'Loughlin, 2012). Except for the atlas and axis, each vertebra consists of common structures (Figure 1). Each has a body in the anterior region that is osseous and acts as a weight bearing structure (McKinley & O'Loughlin, 2012). The vertebral foramen sits posterior to the body and acts as an opening to house the spinal cord (McKinley & O'Loughlin, 2012). The rim of the vertebral foramen is known as the vertebral arch and is made up of two laminae, which form the posterior rim of the arch, and two pedicles, which make up the anterior rim of the arch (Gray, 2008; McKinley & O'Loughlin, 2012). The spinous process projects posteriorly from the laminae and two transverse processes project laterally from the sides of the vertebral arch (Gray, 2008; McKinley &

O'Loughlin, 2012). Each vertebra also has superior and inferior articulating surfaces as well as facets (smooth articulating surfaces) that change in orientation depending on the region of the vertebral column they are in (McKinley & O'Loughlin, 2012). Each vertebra (except C1 and C2) has a fibrocartilaginous disc between them that mainly acts to resist compression and varies in thickness by region of the spine (Banerjee et al., 2004; Goel et al., 1984; Gray, 2008; McKinley & O'Loughlin, 2012).

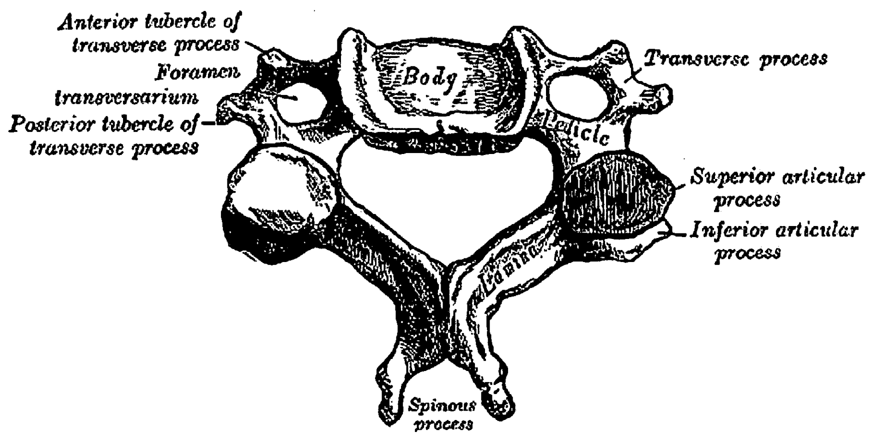

Figure 2.1.1.1 Typical characteristics of a cervical vertebra (From Gray, H. (2008). Anatomy descriptive and surgical. (Second ed., p. 8). London, England: Arcturus). Adapted with permission from Arcturus: [Anatomy Descriptive and surgical] (See Appendix B), copyright (2008).

For the purpose of this study, a more in depth description on the specific anatomy of the cervical spine (C1 to C7) will now be discussed. The cervical spine is divided into two sections: the upper cervical spine (C1and C2) and the lower (C3 to C7) (Gray, 2008). Although this study will not directly be measuring and monitoring motion at the C1 and C2 levels, it is necessary to be aware that injuries to these structures can be related to SCI due to the intimate connection of the spinal column (see Section 2.2, p. 12). It is however, important to discuss C3 to C7 level and their surrounding structures in more detail as direct measures at these levels will be taken throughout this study. The C3 to C7 vertebrae are similar in shape, articulation and orientation. The C7 vertebra has the most distinct characteristics as is marks the transition from the cervical to the thoracic spine. The most distinguishing feature is the long slender spinous process (Gray, 2008). Motions that occur in the lower cervical spine include neck flexion, extension, lateral flexion and rotation and the contribution of a specific spinal segment to that motion varies by level (Banerjee et al., 2004; Goel et al., 1984). Flexion begins in the lower cervical spine, then motion occurs from the level of the occiput down the spine to C4 followed by a brief period of extension at C6 and C7 (Swartz et al., 2005). The vertebrae are connected and motion is controlled primarily by the soft tissue of the cervical spine which consists of ligaments, capsules, discs and muscles (Goel et al, 1984). Knowledge of the bony features of the cervical spine are directly related to this study as the majority of injuries that lead to catastrophic spinal injuries involve not only the vertebrae as a complete structure, but its separate features as well. Injuries related to the osseous anatomy of the cervical spine will be discussed further in Section 2.2 (p.12).

2.1.2 Soft Tissue of the Cervical Spine

Motion in the cervical spine occurs as a result of the surrounding musculature acting on it, but this motion is governed by the ligaments (Goel et al., 1984). Although the soft tissue acts together to produce movement, it also limits it and at times play a crucial role in maintaining the integrity of the cervical spine (Yoganandan, N., Kumaresan, S. & Pintar, F., 2001). Ligaments, composed of collagen and elastin, vary in structure and attachment based on the level of cervical vertebrae (Yoganandan et al., 2001). These unique characteristics have a direct influence on each of the ligament's specific structure and function (Siegmund et al., 2009). They connect the vertebra and act to resist shearing and distractive forces depending on the external load applied to them (Yoganandan et al., 2001). The internal response that they exhibit is then based on the mechanical properties and anatomical structure of the ligaments themselves (Yoganandan et al., 2001). They have been shown to absorb energy during high speed impact as well as provide passive stability to the neck (Siegmund et al., 2009). The ligamentum nuchae is a cord-like band that runs from the greater occipital protuberance to the C7 vertebra attaching the tips of the spinous processes (Goel et al., 1984). The supraspinous ligament originates at the ligamentum nuchae and continues along the tips of the spinous processes to the sacrum (Goel et al., 1984; McKinley & O'Loughlin, 2012). The interspinous ligaments run from the root to the apex of the spinous processes and connect adjacent processes (Goel et al., 1984; Gray, 2008; McKinley & O'Loughlin, 2012). Both the supraspinous and interspinous ligaments resist flexion in the cervical spine. The ligamentum flavum are broad ligaments that connect the posterior superior rim of the laminae below to the posterior inferior laminae of the vertebrae above (Goel et al., 1984; Gray, 2008; McKinley & O'Loughlin, 2012). Intertransverse ligaments connect adjacent transverse processes and acts to resist lateral flexion (Gray, 2008). There are also capsular ligaments that surround each of the facet joints (Goel et al., 1984). The anterior longitudinal ligament connects the anterior surfaces of the bodies of vertebrae from axis to the sacrum and resists extension in the cervical spine (Gray, 2008). The posterior longitudinal ligament connects the posterior surfaces of the bodies and also runs from the axis to the sacrum (Gray, 2008; Figure 2). It acts to resist flexion. Ligaments in the cervical spine region are designed to resist tensile or distractive forces. Ligamentous disruption during injury can result in an unstable spine which is defined as more than 3.5mm displacement of one vertebra on another in a horizontal direction (Swartz et al., 2005). Swartz et al. (2005) note that when faced with a potential SCI, first responders must approach and treat the victim as the worst case scenario, being that of an unstable spine, because measuring instability in an emergency scenario is not possible nor practical. It is important to understand the function of the ligamentous system in the cervical spine for this study as the ligaments have a direct influence on the severity of the injury as well as the boarding process. This is because the ligaments will be involved in additional stabilization of the neck during patient transfer.

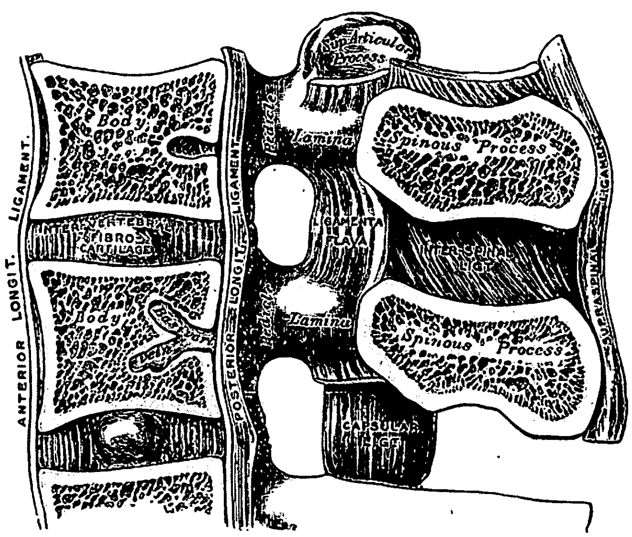

Figure 2.1.2.2 Lateral view of ligaments of the cervical spine up to C3 (From Gray, H. (2008). Anatomy descriptive and surgical. (Second ed., p. 126). London, England: Arcturus). Adapted with permission from Arcturus: [Anatomy Descriptive and surgical] (See Appendix B), copyright (2008).

Muscles play a role in movement as well as stabilization of the spine (Cusick & Yoganandan, 2002). In terms of size, the majority of volume of a neck is composed of the musculature because over 20 muscles surround the cervical spine in an intricate pattern (Siegmund et al., 2009). There are both superficial muscles, such as the sternocleidomastoid (SCM), that attach to the occiput, shoulder girdle and ligamentum nuchae, but do not directly attach to the cervical vertebrae themselves (Siegmund et al., 2009). They still have an effect on cervical motion because they influence movement at the head which results in simultaneous motion transferred down the neck. The deep layers of muscles, such as the interspinalis and multifidus, have several attachment sites directly on the vertebrae (Falla, et al., 2002; Siegmund et al., 2009; Sommerich et al, 2000). These deep muscles are responsible for directly controlling vertebral position (Cusick & Yoganandan, 2002; Falla, et al., 2002; Siegmund et al., 2009; Sommerich et al., 2000). SCM, trapezius, splenius capitus and semispinalis muscles are important for maintaining head stabilization against externally applied loads (Falla et al., 2002; Sommerich et al., 2000). The majority of muscles throughout the neck have vertically oriented fibres causing increased axial compression to the spine when activated (Siegmund et al., 2009). Their structure is composed of high density muscle spindles (Siegmund et al., 2009) and although they are elastic in nature, they will resist motion even when not active. Muscle activation is thought to play a role in the exacerbation of certain injuries including indirect strain on other anatomical structures in the neck (Cusick & Yoganandan, 2002; Siegmund et al., 2009). Neck muscle activation results in a change in head and neck kinematics, thus thresholds related to injury and load may be exceeded (Siegmund et al., 2009). Knowledge of the soft tissue of the spine is necessary for this study. It plays an integral role not only in failing to aid in protecting against catastrophic injury, but it can also affect the boarding process itself either through instability, caused by primarily by ligament damage or muscle spasm shifting unstable structures and leading to exacerbation of injury. Likewise, the muscle activity from SCM, upper trapezius and the upper erector spinae muscles will be monitored and recorded during the collection process. This is discussed further throughout Section 3 (p. 29).

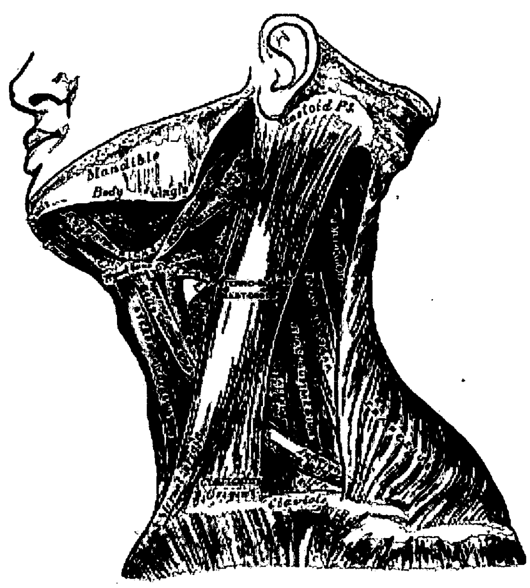

Figure 2.1.2.3 Musculature of the cervical spine (lateral view) (From Gray, H. (2008). Anatomy descriptive and surgical. (Second ed., p.180). London, England: Arcturus). Adapted with permission from Arcturus: [Anatomy Descriptive and surgical] (See Appendix B), copyright (2008).

2.1.3 Spinal Canal and Spinal Cord

The spinal cord travels inferiorly from the foramen magnum, through the vertebral foramen to the L1/L2 disc level (Gray, 2008). It is protected by the ligamentous elements that line this foramen (Banerjee et al., 2004). The spinal cord occupies approximately 50% of the spinal canal at the level of C1 and 75% of the area in this canal in the lower cervical spine (Banerjee et al., 2004; Cusick & Yoganandan, 2002). The cervical spine is lengthened in flexion and shortened in extension resulting in a corresponding change in the spinal canal dimensions (Cusick & Yoganandan, 2002). The spinal cord itself also lengthens in flexion and shortens in extension (Cusick & Yoganandan, 2002). Ideally, there is a 1:1 spinal canal-vertebral body relationship (Chao et al., 2010). There have been reports that a canal-body relationship smaller than 1:1 makes individuals more susceptible to not only spinal cord injury in general, but should injury occur, the severity of the injury will be increased (Chao et al., 2010). During extension, infolding of the ligamentum flavum along with laminar movement will cause a significant reduction in the cross sectional diameter of the canal (Cusick & Yoganandan, 2002). The spinal cord itself undergoes distortions with functional positioning and although effects are primarily seen at the site of deformation, because it is a continuous structure, effects may also be seen more distally along the tract (Cusick & Yoganandan, 2002). In flexion, the cord lengthens and compresses in the anterior-posterior diameter and reverses during extension (Cusick & Yoganandan, 2002). The spinal cord and the canal are two of the most significant structures related to the study. One of the goals of the study is to help improve the management of SCI therefore, knowledge of the structure and function of the spinal cord and canal themselves is important.

2.1.4 Anatomy Summary

The anatomy of the cervical spine is intricate and complex making it a unique functioning unit. The bones are oriented and articulate in a manner that allow for various movements which are driven by muscle activation and external forces. SCM and trapezius muscles are two superficial muscles that are important in stabilizing the head and neck. As the vertebrae go through various ranges of motion, the amount of available space there is for the spinal cord to travel through the canal changes. The spinal cord itself will undergo changes in cross sectional area during neck movement. Based on its location within the body the neck and its surrounding structures are prone to injury. The following section describes the main mechanisms related to spinal injury.

2.2 Mechanisms of Cervical Spine Injury

There are several factors that decrease space in the spinal canal or decrease load tolerance and therefore render individuals at risk for SCI (Chao et al., 2010; Cusick & Yoganandan, 2002). These include variables such as age, sex, degenerative diseases and congenital anomalies (Cusick & Yoganandan, 2002). Participation in contact sports is also a major risk factor for sustaining a SCI. Sports participation ranks fourth among the list of causes of SCI (behind motor vehicle accidents, violence and falls) and in the first three decades of life, it ranks second overall (Banerjee et al., 2004). The highest injury rates occur in men's football, women's gymnastics and men's hockey (Banerjee et al., 2004).

There has been much debate over the classification of injury related to the spine due to the many factors contributing to its definition (Cusick & Yoganandan, 2002; Taylor & Taylor, 1996). These factors mostly include interrelationships between biomechanical influences (i.e. magnitude, rate and direction of force applied) (Cusick & Yoganandan, 2002; Taylor & Taylor, 1996). In order to classify the threat of neurologic involvement, the degree of instability as well biomechanical factors must be noted (Cusick & Yoganandan, 2002; Taylor & Taylor, 1996). Bone, intervetebral discs and ligaments undergo increased stiffness with increased rate of loading, but this effect is more substantial on bone (Cusick & Yoganandan, 2002). This study is directly related to the biomechanical factors related to injury, the skill of the first responder and how both of these affect the boarding process.

The anatomy of the cervical region is designed so that forces are absorbed and dissipated through musculature and discs in the spine's natural resting position in which the cervical spine is placed slightly in extension (Banerjee et al., 2004; Chao et al., 2010). When the spine is at 30° of flexion however, it acts as a rigid column and when axial compressive loads are encountered, the forces are sent directly to the vertebrae. Thus, this force causes the spine to buckle, resulting in injury including fracture, dislocation or subluxation (Banerjee et al., 2004; Chao et al., 2010). The effect of axial compressive loading on the spine column is illustrated in Figure 2.2.4.

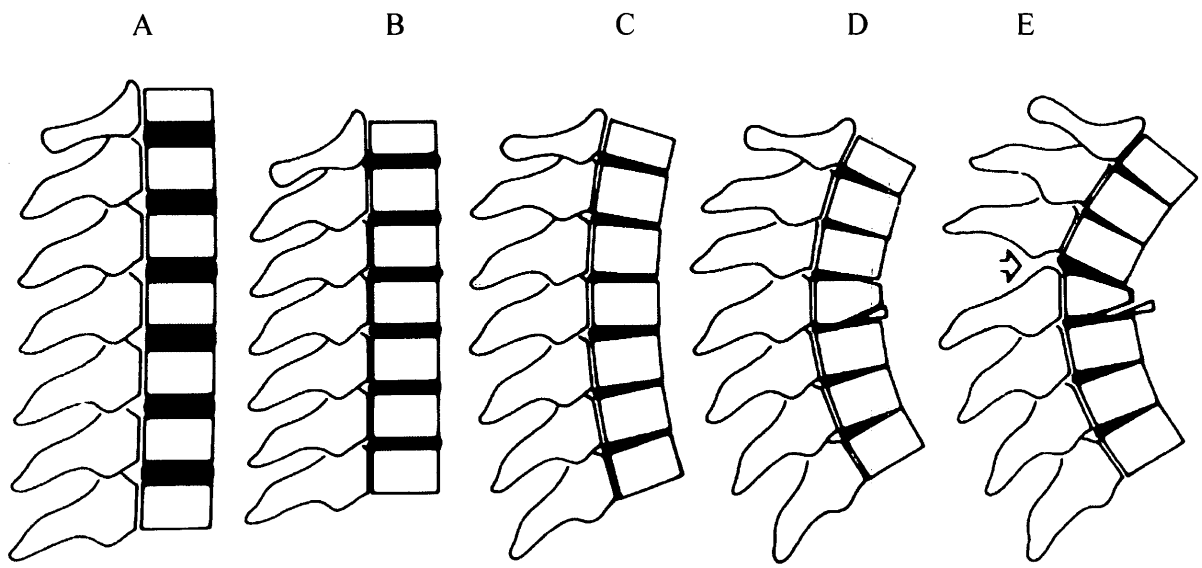

Figure 2.2.4 The buckling effect of the spine. *A)* The spine acts as a solid column at 30° of flexion. *B) and C)* As axial compressive loads are applied the spine begins to buckle resulting in injury *(D and E)* (From Chao, S., Pacella, M. J., & Torg, J. S. (2010). *Sports Medicine), 40*(1), 59-75, p. 61, Fig. 1). Adapted with permission from Springer : [Sports Medicine] (See Appendix C), copyright (2010).

Between 1971 and 1976, it was widely accepted that axial loading caused the most significant injuries in football with 85% resulting in paraplegia (Chao et al., 2010). Banerjee et al. (2004) noted that fracture/dislocations (that occur as a result of this axial loading on the spine) accounted for up to 80% of catastrophic spinal cord injuries in football from 1977 to 2001. The introduction of the no spearing tackling rule by the National Collegiate Athletic Association (NCAA) implemented in 1976 reduced the incidence of 34 catastrophic injuries in 1975, to eight reported cases in 2007 (Chao et al., 2010), a reduction of 425%. Chao et al. (2010) report that up until 1988 axial loading was the primary mechanism of injuries occurring at the C3-C4 level (which are rare and often unreported). These include injuries such as bilateral facet dislocation, acute disc herniation and C4 vertebral fracture (Chao et al., 2000). This renders the spine unstable and with an axial load, irreversible SCI with permanent quadriplegia typically results (Chao et al., 2010).

2.2.1 Axial Loading Injuries

Axial loading causes two types of fractures most commonly associated with SCI (Banerjee et al., 2004). The first type are known as burst fractures and result in shattered vertebrae where fragments disperse in all directions, including towards the spinal cord (Banerjee et al., 2004). These fractures have the most potential for spinal canal compromise (Banerjee et al., 2004; Cusick & Yoganandan, 2002). The second fracture type occurs with flexion combined with compression and is known as a teardrop fracture. It is characterized by a shortened anterior column (as the vertebral body fails under compression) and a failure of the spinal ligaments with a lengthened posterior column (Banerjee et al., 2004). Most fractures occur in the lower cervical region due to the increased lever effect of the upper cervical spine (Banerjee et al., 2004; Cusick & Yoganandan, 2002). In a study examining 109 postmordem spines exposed to blunt force trauma, the second most common fracture site (after the C2 level) was the C6 level. Although SCI to the upper cervical spine is rare, it does not usually result in significant neurologic impairment because of the large amount of space available for the spinal cord, as compared to the lower cervical spine (Banerjee et al., 2004; Cusick & Yoganandan, 2002). In fact, injuries such as the Hangman fracture (traumatic spondylolisthesis of the axis) and Jefferson fracture (a burst fracture of the atlas) actually expand the spinal canal (Banerjee et al., 2004). Those injuries that cause instability of the atlantoaxial complex, such as odontoid fracture and transverse ligament rupture, will result in neurologic defects and may include diaphragmatic paralysis and respiratory distress (Banerjee et al., 2004).

Along with fractures, bilateral facet dislocation is seen with an axial loading mechanism (Ivancic et al., 2007). This results in complete spinal cord lesion and quadriplegia in up to 85% of reported cases (Chao et al., 2010; Ivancic et al., 2007). During bilateral facet dislocation, inferior articular facets of the upper vertebrae shift anteriorly and rotate with respect to the superior articulating facets of the vertebrae below, causing damage to the interspinous ligament, facet capsules, the intervertebral disc and ligamentum flavum (Ivancic et al., 2007; Vacarro et al., 2001). There are also reports of the posterior longitudinal ligament being damaged as a result of this injury (Vaccaro et al., 2001).

2.2.2 Hyperflexion and Hyperextension Injuries

Hyperflexion and hyperextension more recently have been documented as mechanisms of injury that cause the most severe cervical spine injuries and are also primary contributors to injuries resulting in cord deformation, contrary to axial loading (Chao et al., 2010). The spine tends to be more prone to extension type injuries because the mass of the posterior neck musculature outweighs that of the anterior (Taylor & Taylor, 1996). When the cervical spine is exposed to flexion combined with axial compression, the spine bends such that the anterior column is shortened while the posterior column is lengthened. The spinal cord may get pinched in between the vertebral bodies (Banerjee et al., 2004; Cusick & Yoganandan, 2002; Chao et al., 2010).

Extension or flexion combined with compression can also lead to fractures such as unstable "flexion teardrop fracture" that is often associated with SCI (Banerjee et al., 2004; Cusick & Yoganandan, 2002; Chao et al., 2010). The odontoid process may also fracture with flexion or extension (Cusick & Yoganandan, 2002). These types of fractures account for one quarter of all fractures that occur in the cervical spine though they are not often associated with neurologic deficit (Cusick & Yoganandan, 2002). Hyperflexion and hyperextension forces also have a tendency to cause ligamentous disruption (Chao et al., 2010; Cusick & Yoganandan, 2002).

2.2.3 Lateral Flexion and Rotation Injuries

Lateral flexion and/or rotation injuries to the cervical spine do not generally result in catastrophic SCI. Conditions such as "burners" or "stingers" are caused by forced neck extension combined with rotation away from the affected side and result in unilateral, transient neurologic symptoms (Banerjee et al., 2005). Similarly, if the shoulder is forced into depression and the head is forced into lateral flexion away from the depressed shoulder, tensile forces involved can cause temporary neurologic symptoms (Banerjee et al., 2005). Symptoms associated with these injuries include unilateral paresthesia, radiating pain and weakness on the affected side (Banerjee et al., 2005). These symptoms typically resolve within 24 to 48 hours (Banerjee et al., 2005). Along with these neurologic symptoms, the athlete will present with pain free neck range of motion and no midline tenderness. These types of injuries are fairly common in collision type sports, but it is necessary to be aware of the mechanism and symptoms associated with lateral flexion/rotation injuries in order to rule out catastrophic SCI and manage them appropriately.

2.2.4 Mechanisms of Cervical Spine Injury Summary

There are several mechanisms leading to cervical spine injury ranging from predisposing congenital anomalies to traumatic directional forces. Of these, forced hyperflexion is the leading cause of catastrophic cervical spine injury during sporting activities involving contact (Chao et al, 2010). Congenital spinal anomalies, unstable fracture/dislocation and acute central disc herniation result in catastrophic SCI and are associated with neurologic involvement. It is necessary to be aware of the mechanisms related SCI in order to provide effective management of these injuries.

2.3 Current Patient Transfer Methods

Currently, health care practitioners involved in the initial management stages of SCI, use similar criteria to determine if there is a catastrophic cervical spine injury requiring further radiologic evaluation. Athletic therapists use the following criteria: altered level of consciousness, bilateral or unilateral neurologic findings, and/or significant midline spine pain with or without palpation and/or spine deformity (Swartz & Del Rossi, 2009). Canadian paramedics and nurses use the Canadian C-Spine Rule which involves the use of three high-risk criteria, five low-risk criteria and the ability of patients to laterally rotate their neck to determine if the injured individual requires more in-depth evaluation (Vaillancourt et al., 2009). Regardless of the criteria used to evaluate the severity of the SCI, immediately following the injury, the spine must be handled as if it is unstable until further evaluation (radiologic) can be performed (Conrad et al., 2012).

On field, if a serious cervical spine injury is suspected, the patient is immobilized and prepared for transport to the hospital. This includes aligning the spine to a neutral position, applying a cervical collar and securing the patient onto a long spine board. Some authors suggest splinting the head and neck in the position found while others favour moving the neck into "neutral position" to minimize movement which reduces spinal cord morbidity and facilitates airway management (DeLorenzo et al, 1996, Swartz et al, 2009). However, there is no clear definition as to what "neutral" position is, whether or not this is actually the most optimal position for the unstable injured spine or when to do this alignment (DeLorenzo et al, 1996). Contraindications to neck realignment include creptitus, spasm, increased pain or neurological symptoms (Swartz, 2009). Provided the airway is not compromised in any way, the patient is supine lying and the neck is in "neutral" position, a cervical collar is applied to the patient's neck to help maintain stabilization prior to transfer onto a long spine board. Recently evidence has suggested that applying a cervical collar may also invoke too much motion. However, for the purpose of this study, collar application will not be addressed.

After collar application, the next step is to place the athlete onto a spine board. Special consideration must be taken into account for those athletes wearing protective equipment. The need for equipment removal is dependent on a number of factors including the vital sign stability of the athlete. For the purpose of this study, equipment removal will not be included or addressed, but is suggested for future investigation (see Section 8, p. 68).

There are two main methods for moving a supine individual onto a spine board: the log roll and the lift-and-slide technique. The log roll technique (Figure 2.3.5, p. 19) involves five individuals: one providing manual head-neck stabilization; three individuals assisting in rolling the torso (at shoulder), upper and lower extremities (at hip and legs); and one individual controlling the spine board (Conrad et al., 2012; Del Rossi et al., 2004). The individuals performing the roll kneel on the board that is placed lengthwise against the injured athlete on the side opposite to the direction the head is facing. The individuals then roll the body towards them onto the long spine board into a supine position.

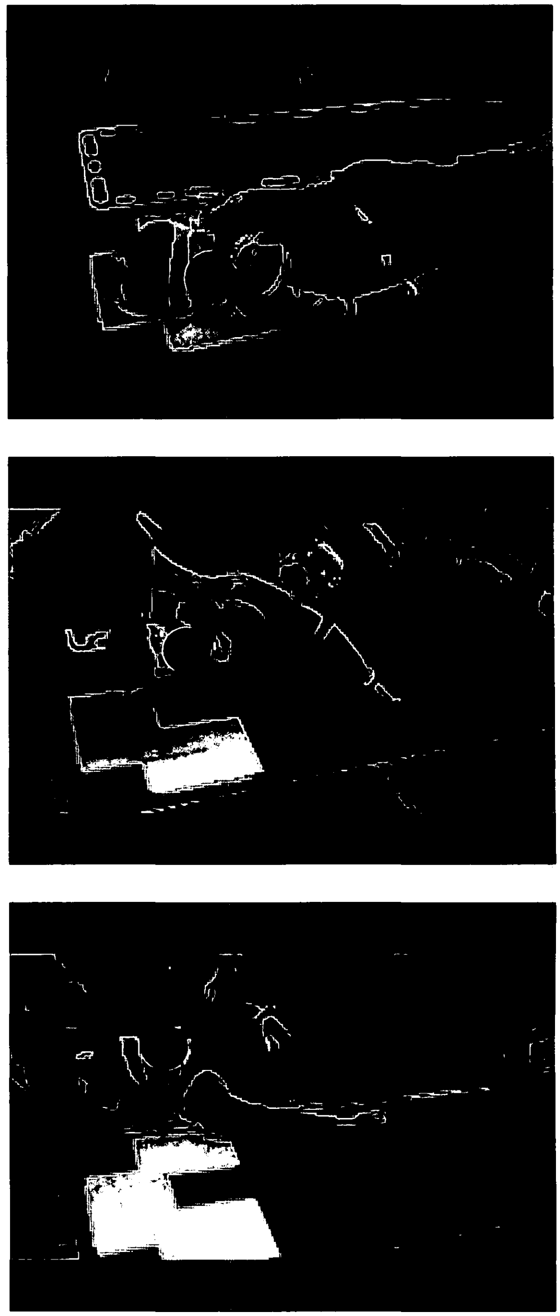

Figure 2.3.5 The log roll manoeuver from a prone starting position. *Top* Starting position for log roll. *Middle* Transition roll onto the spine board. *Bottom* Final supine position.

The second transfer method is the lift-and-slide technique (Figure 2.3.6). There are several configurations of first responders involved in this technique. For example, one person is responsible for maintaining inline manual stabilization of the head-neck while four people (two at the upper extremities kneeling and two at the lower extremities straddling the injured) lift the victim onto a spine board that is placed under the body by a sixth individual (Conrad et al., 2012; Del Rossi et al., 2004).

Figure 2.3.6 The lift-and-slide technique (From Del Rossi, G., Horodyski, M., Heffernan, T. P., Powers, M. E., Siders, R., and Brunt, D. (2004). *Spine, 29*(7), E 134-8, E 135, Fig. 2). Adapted with permission from Lippincott Williams and Wilkins/Wolters Kluwer Health: [Spine] (See Appendix D), copyright (2004).

The National Athletic Trainers Association position statement suggests that when an athlete is lying supine the lift-and-slide technique should be used and when the athlete is prone the log roll is the appropriate method (Swartz, 2009). Various studies have suggested that the lift and slide technique is a superior method to the log roll because it causes less axial rotation, lateral flexion and lateral translation of the spine (Swartz, 2009). The most recent research strongly advises against the use of the log roll technique for supine cases (Conrad et al., 2012). However, the log roll remains the preferred method of transfer because there are less people needed to perform this technique effectively as compared to the other transfer methods (Del Rossi et al., 2009). Conrad et al. (2012) suggest several reasons for why the log roll technique is preferred such as complacency (causing a false sense of assurance in the technique) and poor risk assessment as secondary neurologic injury is difficult to assess without a proper baseline. Even though some movement is inevitable, it is currently unknown how much movement is acceptable without causing harm using either technique (Conrad et al., 2012; Swartz et al., 2009). Therefore, regardless of the technique chosen, the goal is to minimize as much motion as possible. Once the athlete is safely placed on the spine board and secured to it, transport to the hospital for further evaluation is arranged.

2.3.1 Patient Transfer Methods Summary

Health practitioners use a variety of on field classification systems to determine whether an injured individual has a suspected spinal cord injury and requires further evaluation. During the immobilization process of a supine patient, the spine is aligned to neutral position, a collar is applied and the injured is moved onto a spine board. Literature has shown that the log roll technique causes more motion than other transfer technique and is strongly advised against. However, at the present time it is the only technique used during prone patient transfer because it allows for quick access to the airway and will therefore be used during this study.

2.4 Methodology Literature

2.4.1 Kinematics

2.4.1.1 Optoelectronic Motion Analysis Systems

Movement of the cervical spine can be difficult to measure because of the complex structure and movements located in this region of the body. The method by which a researcher chooses to measure this motion is therefore based on the specific goals of his or her study. This study captured 3D movements occurring at the cervical spine using accelerometers (described in Section 2.4.1.2, p.25). To capture motion occurring in 3D space an inertial reference system is used to relate a body's position and its orientation in space as defined by bony landmarks (Cappozzo et al., 1995; Cappozzo et al., 1997; Giasanti et al., 2003). Passive based motion analysis systems use infrared reflective markers placed on the bony landmarks on the participant to define a segment. The markers emit a signal frequency which is picked up by cameras and represented by a two-dimensional image. The bony axes that are outlined by the reflective markers are then related to a global coordinate system to convert this into 3D data (Cappozzo et al., 1997). There must be at least two cameras present to capture each marker and calculate the segment's coordinate in space (Baker, 1997). The systems that use video cameras capture motion at a high sampling rate so are ideal for recording 3D movement (Wong et al., 2007). There are several types of 3D motion capture analysis systems, such as Optotrak® Certus™ and (Northern Digital Inc., Waterloo, Ontario, Canada) and Vicon MX (Vicon Motion Systems, Colorado, USA) that are popular in both research and clinical laboratories and produce both reliable and reproducible methods (Wong et al., 2007). These motion capture systems tend to represent the gold standard for 3D motion.

However, they are expensive, can be time-consuming to set up and are typically restricted to confined laboratory settings (Giasanti et al., 2003; Wong et al., 2007). Also, their reliability depends on positioning of cameras in order to capture the reflective markers as these markers can become easily hidden resulting in incomplete data (Giasanti et al., 2003). An increasingly popular alternative to these 3D motion capture systems is the use of inertial measurement units.

2.4.1.2 Inertial Measurement Units

Accelerometers are inexpensive small force transducers used to measure tilt, 3D angles, acceleration, and angular velocity (Winter, 2005; Wong & Wong, 2008; XSens Technologies B.V. ©, 2008). Based on these measures, information on position and orientation in space can be obtained regardless of the position of the accelerometer itself (Wong & Wong, 2008; Wong et al., 2007). A single accelerometer will measure linear accelerations. However, in order to obtain 3D information, three linear accelerometers are mounted at right angles to each other within the same unit (Winter, 2005). This is known as a tri-axial accelerometer. The use of a single or biaxial accelerometer can produce error based on limb position in arbitrary directions and are limited to low frequency ranges, making tri-axial accelerometers ideal for dynamic situations (Hanssen et al., 2006).

Gyroscopes are units that contain vibrating parts used to measure angular velocities (Woodman, 2007). There are many varieties of gyroscopes including mechanical and optical. In micro-electro-mechanical systems (MEMS), which are used extensively in research today, gyroscopes measure the Coriolis effect. This effect involves force developing from a rotating reference frame (Wong et al., 2007). A mass sends a vibration down an axis and when the gyroscope is rotated, a secondary vibration is sent down a perpendicular axis based on the Coriolis effect and as a result the angular velocity can be calculated (Woodman, 2007).

An inertial measurement unit has three orthogonal rate gyroscopes combined with a tri-axial accelerometer (Woodman, 2007). Some of these units (including those being used in this study) also contain a magnetometer, which provides a stable reference as it measures gravity and the earth's magnetic north. The additional information provided from magnetometers helps to eliminate drift caused by gyroscopes (Wong et al., 2007). These devices have the capability to monitor the absolute orientation of an object as well as the motion of an object in 3D (Woodman, 2007; Jasiewicz et al., 2007). These units use the integrated signal of the gyroscope (which creates the known orientation) to determine the global coordinates and track position of the accelerometer signals (Woodman, 2007). Also, acceleration, velocity and displacement measures can be obtained as the transducers collect data individually (Jasiewicz et al., 2007). Several studies have suggested that these units are comparable in terms of accuracy to the gold standard motion capture systems (Giasanti et al., 2003; Jasiewicz et al., 2007). They have the distinct advantage of being lightweight, portable, inexpensive and they allow participants to move without hindrance of cabling (Giasanti et al., 2003; Jasiewicz et al., 2007). MEMS also have the advantage of data integrity as the effectiveness of motion capture systems relies on camera placement and data is lost with marker obstruction. These units can also be used in conjunction with other systems such as electromyography without experiencing interference (Jasiewicz et al., 2007).

2.4.1.3 Kinematics Summary

Accelerometers and gyroscopes are transducers used to measure variables such as 3D angles, acceleration, displacement and angular velocity. When combined with magnetometers, they are known as inertial measurement units and can provide valuable information regarding a segment's location in 3D space. Although the gold standard for 3D motion research tends to be captured by optoelectronic equipment, there are disadvantages to their use such as cost and loss of data due to poor camera: sensor relativity. MEMs on the other hand, are small, portable, and easy to use, making them an ideal alternative to optoelectronic motion capture systems. A seven camera Vicon optoelectronic system was available, but due to the number of participants and the nature of this study, data would have been severely incomplete and unusable. Therefore, XSens was the most ideal choice for maintaining integrity of data (see Section 4.4.2, p.37).

2.4.2 Electromyography

EMG is a technique used to record and analyze electrical signals from active motor units within a muscle, known as motor unit action potentials (MUAPs). The resting membrane potential of a muscle is -70mV. During a change in the muscle fiber membrane's permeability, the fiber becomes activated and the membrane depolarizes to approximately +20mV. This wave, known as an action potential, is controlled by motor neurons. The action potential travels to the motor unit endplates. Electrochemical changes then occur at the endplate resulting in a stimulus being sent down the transverse tubules in a muscle fiber. This, in turn, causes a release of calcium ions (Ca^{2+}) into the sarcoplasmic reticulum (Winter, 2005). This depolarization waveform then propagates along the muscle towards the tendons from a site in the muscle known as the innervation zone (Merlo & Campanini, 2010). This causes a field of electrical voltage to be generated which is detectable at the skin (Merlo & Campanini, 2010). EMG is a representation of this change in voltage (depolarization and repolarization) as the electrodes record a summation of all active MUAPs in a muscle fiber within the detection area of the electrodes (Winter, 2005).

There are many factors that influence the muscle activation signal including velocity of muscle shortening or lengthening, reflex activity, muscle temperature, changes in length of a muscle during dynamic contraction, etc. (Winter, 2005). These affect the amplitude of the signal as well as the number of active motor units, the muscle fiber diameter, the depth and location of the innervation zone with respect to the electrode detection surfaces and conduction velocity of the action potentials (DeLuca, 1997; Merlo & Campanini, 2010). Tissue impedance or the impedance of the transmission of the electrical activity at the muscle level depends on the type and amount of tissue between the detection surface and the muscle fiber can alter the signal itself because it has an effect on the conduction velocity (DeLuca, 1997).

By analyzing the raw and processed waves, valuable information can be gained from the use of EMG. Raw signals can provide information on coarse amplitude and initiation of muscle activity, but are not useful for comparing different muscles across different trials. Interpreting EMG signals involves critically investigating the following: levels of muscle activity (by looking at the magnitude of amplitude as well as timing of activation), activation characteristics (i.e., asymmetries in amplitude and changes in co-activation), and various kinematic and postural changes for gait analysis and rehabilitation purposes. Signals that are processed provide valuable information on signal amplitude and frequency. For example, obtaining the linear envelope of the signal, through full wave rectification and then applying a low pass filter, gives an indication of the amount of activity within a muscle as well as the onset of muscle activity (Robertson et al., 2004). Integrated EMG sums the amount of activity over a period of time and is useful in quantifying EMG relationships (Roberson et al., 2004). It is also used an indicator of muscular effort. Another way to analyze EMG data is to look at frequency content of the signal (Robertson et al., 2004). This includes determining the mean and median frequency as indicators of changes in muscle conduction velocity as well the onset of fatigue (Robertson et al., 2004).

EMG is also affected by the type electrodes. There are two types of EMG techniques: surface EMG (sEMG) and indwelling or needle EMG (nEMG). sEMG uses electrodes placed superficially to give a gross representation of electrophysiological muscle activity (Soderberg & Cook, 1984; Haig et al., 1996) as it is capable of recording from numerous active motor units as it has a larger detection area. sEMG has a narrow frequency (20 to 500Hz) and low-signal resolution (Pullman et al., 2000). Alternatively, nEMG uses a fine wire embedded in the muscle to evaluate neuromuscular diseases and gait disorders (Pullman et al, 2000) and has a larger frequency band (20-1000Hz) (Winter, 2005) . nEMG detects activity of single motor unit in deep muscles, but there are still no definitive guidelines for their placement within a muscle belly (Soderberg & Cook, 1984; Haig et al., 1996). Although nEMG serves a distinct purpose, it is invasive, there is risk of wire fracture, and its sample area may not be indicative of the activity within the whole muscle (Soderberg & Cook, 1984). It is necessary for the researchers/evaluators to have a clear understanding of variables being measured in order to ensure the best EMG technique is applied. This study will use surface electromyography (sEMG) in particular because it provides a non-invasive method for analysing muscle activity and gross movement (Falla et al., 2002). Therefore, the remaining EMG discussion will focus on sEMG.

The placement, orientation and size of surface electrodes have an effect on the signal characteristics. sEMG electrode placement is usually described based on location over the muscle belly, away from the neural end point muscle, and parallel to the muscle fibres because these affect the frequency and amplitude characteristics of the signal (DeLuca, 1997; Falla et al., 2002; Zipp, 1982). In a study by Falla et al. (2002) it was shown that in order to increase accuracy and reliability of sEMG, electrodes must be placed over the lower portion of each muscle because the midpoint or upper portion of muscles showed inconsistencies in sEMG values. The size of the electrode itself will capture motor unit activity within its direct vicinity (DeLuca, 1997). Therefore, a larger detection area will increase the amplitude of the signal as more motor unit activity will be detected, but the frequency content of the signal is decreased (DeLuca, 1997). With regards to the distance between electrodes and the size of electrodes themselves, Zipp (1982) advises the use of large electrodes with large interelectrode spacing on large muscles and small electrodes with narrow interelectrode spacing to reduce cross-talk signals from adjacent muscles. This spacing determines the bandwidth of the signal (DeLuca, 1997). As the spacing increases, so does the detection area, however there is a decrease in the detection depth. The typical interelectrode distance is 4-5cm for large electrodes and 2 to 2.5cm for small (Zipp, 1982).

For the purpose of this study, bilateral sEMG was used to monitor and assess the presence of muscle activity in the SCM, trapezius and cervical erector spinae. It has been shown that these muscles are involved in stabilization of the neck and are most easily accessible by sEMG (Sommerich et al., 2000).

2.4.2.1 Electromyography Summary

EMG is method of recording myoelectric activity and is used in the assessment of muscular fatigue, state of activation (on/off) and overall muscle activation level. The EMG signals represent the changes in voltage that occur during muscle activation. Accuracy and reliability of sEMG are highly influenced by electrode location, position and size of the electrode. It is recommended that electrodes be placed parallel to the muscle fibers and approximately 2 to 2.5cm apart to maintain the fidelity of the signal.

3. Introduction

The purpose of this study was to collect and analyse cervical spine movement during prone patient transfers onto a spine board using the log roll technique in order to determine the correct timing of neck realignment that invokes the least amount of movement during realignment to neutral position. This study also aimed to expand knowledge on safe extrication procedures during spinal injury, specifically the log roll technique for prone victims, in order to reduce overall movement of the neck thus preventing further injury. Current methods of diagnosis and management of cervical spine injuries differ slightly across health care professions. This study focussed specifically on the realignment phase of the head that occurs during the initial management stage of SCI for a prone victim, measuring kinematics and EMG in the neck region. Kinematic measures include changes in angles occurring at the forehead, and the C4, C7 and T4 levels of the spine during several transfer conditions onto a long board. Bilateral EMG of SCM, upper trapezius and cervical erector spinae muscles were also taken to ensure minimal neck muscle activation during the transfer process because this may have an influence on the amount of overall neck motion. These measures were also used to determine the possibility of reflexive neck muscle activity at any part of the log-roll procedure.

Primary response health care practitioners, such as paramedics and Certified Athletic Therapists, perform the primary survey in the same manner. Level of consciousness, airway, breathing and circulation are evaluated. The head and neck are manually stabilized by a healthcare provider until the severity of cervical spine injury can be further evaluated (Blackham & Benger, 2009, Swartz & Del Rossi, 2009). Although there is a variety of research on the initial injury management stage, there is a void with regard to quantifying the amount of cervical spine movement that occurs during prone patient transfer onto a spine board using the log roll technique with respect the timing of head and neck realignment to a neutral position. The following section will discuss the methodology of the study including participant inclusion criteria, data processing and experimental setup.

4. Methodology

4.1 Participants

For this study the following participants were involved: two trained models who acted as victims, 13 Certified Athletic Therapists and 10 student Athletic Therapists. During each data collection, one Certified Athletic Therapist worked with a team of four Athletic Therapists (either students or other Certified Athletic Therapists) and performed log rolls on two different models (one male and one female). The main party of interest was the Certified Athletic Therapist at the head position. A total of 21 sets of data were collected: 8 with the male model and 13 with the female model. Toward the end of the collection period, the male model unexpectedly became sporadically available and another male of similar stature could not be obtained. Four complete sets of data were deemed unusable due to equipment failure. More specifically, within the kinematic trials, data from one sensor during one trial was not usable and in the EMG, the left erector spinae data from seven trials and left and right upper trapezius data from three trials were deemed unusable. Although some data was lost, it is estimated that this will not have a significant effect on the outcome of this study. In order to reduce the number of factors affecting the overall outcome of this study, only female Certified Athletic Therapists were used. Also, during the 13 month recruitment period of this study, no male Certified Athletic Therapists were available, or were not willing to participate.

The models: one male 133.8 kg and one female 56.3 kg, were recruited from the York University undergraduate student population and were used as trained "victims" or suspected spinal cord injury patient models. The Certified Atheltic Therapists and student Athletic Therapists were all current members of the Canadian Athletic Therapists Assocation and all therapists held a valid and current First Responder Health Care Practitioner certification at the time of data collection. Copies of credentials of all participants were retained. Out of the eleven Certified Athletic Therapists, eight therapists had been certified for three or more years and the remaining three had been certified for less than three years (two having been newly certified within the same year as the collection period).

In order to ensure accuracy and safety during the data collection, both the models and the Athletic Therapists (both Certified and students) were free from any neck, back and shoulder pain and for at least 12 months leading up to the collection, and did not seek out medical attention for or had taken days off from work or school due to back, neck or shoulder pain. In addition, the models did not have a history of major injury to the neck, upper back,and shoulder region (e.g. fracture, laceration, etc.). They were also free of any other impairments that would have hindered them from moving normally and comfortably for the length of the collection period.

A description of the study was verbalized to potential participants prior to the study itself. At this time verbal and written consent was obtained from all participants. The protocols and consent forms were approved by York University's Office of Research Ethics Human Participants Review Sub-Committee (Certificate #2011-357, amendment approved 03/29/12).

4.2 Instrumentation

4.2.1 Kinematics

The MTx™ sensors (XSen®, XSens Technologies B.V. ©, Enchede, The Netherlands), seen in Figure 7, contain one 3D gyroscope (to measure the rate of turn), a 3D accelerometer (measuring gravitational acceleration) and a 3D magnetometer (providing earth's magnetic field data as a compass). These sensors were used to provide three degrees of freedom inertial orientation measurements (*MTi and MTx User Manual*, 2008). The sensors were 38mm +/- 0.2 by 21mm by 53mm +/- 0.2 in size and weigh 0.03kg (*MTi and MTx User Manual*, 2008). Combined with the XBus® master system, the sensors communicated with a personal computer via Bluetooth technology. Five sensors were secured to the model in the following locations: one on the forehead, one centered over the spinous process of vertebra C4, one centered over the spinous process of vertebra C7, one on the spinous process of vertebra T4 and one on the sternum.

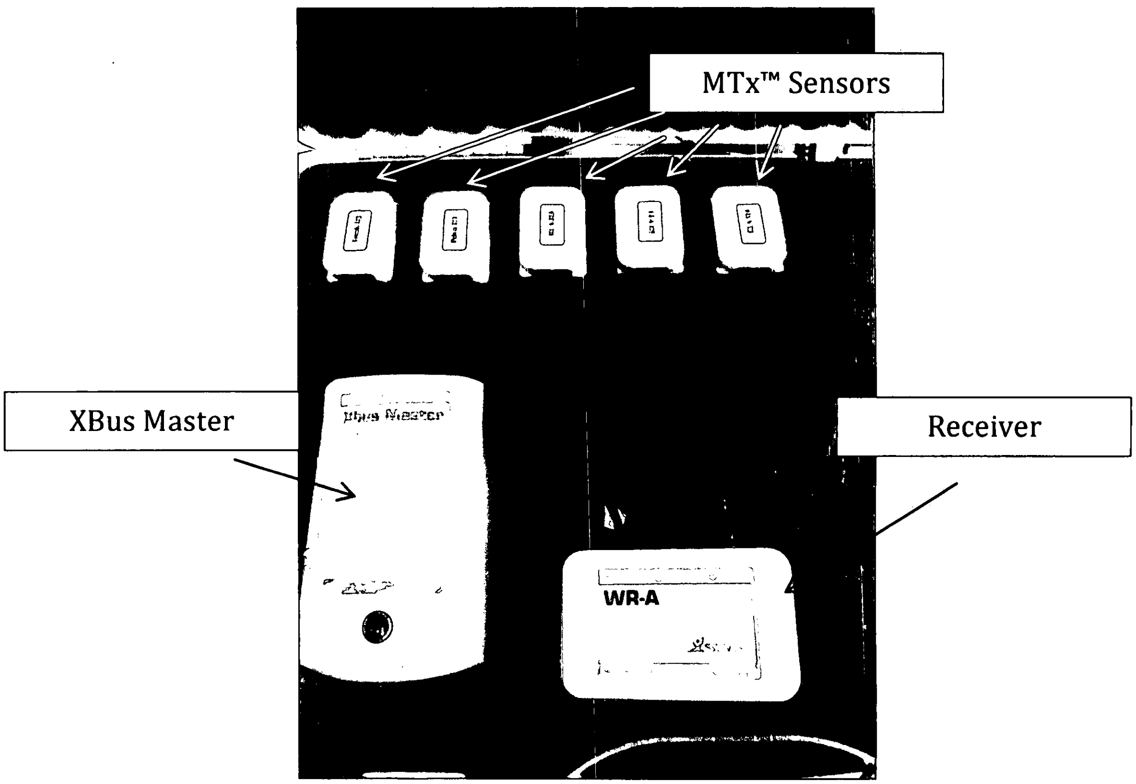

Figure 4.2.1.7 XSens® System including the XBus Master (bottom left) and MTx™ motion sensors (top).

Prior to their application, the skin was cleansed with alcohol swabs (Adams et al., 1986). The sensors were adhered to the skin with Leukotape®. In instances such as the forehead and sensors down the back of the neck, Powerflex® tape was also used to secure the sensors to the model. In order to ensure accuracy and consistency, a measurement from the tip of the nose to the glabella was used to landmark the inferior border of the accelerometer placed on the forehead. For the sternum, the superior edge of the sensor was placed at the sternal notch and due to the size of the sensors, the center of the back side of sensors were placed over the spinous processes of C4, C7 and T4 during each trial. The XBus® system was mounted on the model's left hip in order to ensure it did not interfere with the log roll. Data was sampled at 50Hz. The sensor's orientations were based on Figure 4.2.1.8 below. For this study, rotation (or roll) occurred around the x axis, flexion and extension (pitch) occurred around the y axis and lateral bend (yaw) occurred around the z axis.

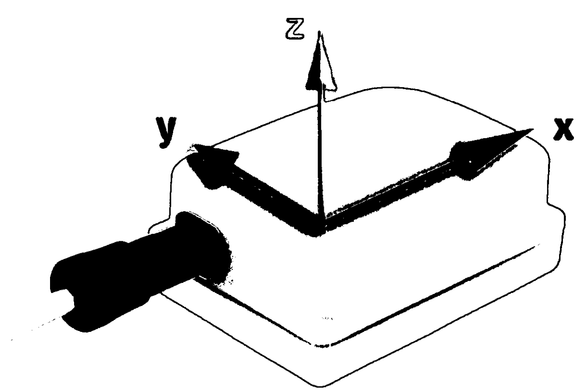

Figure 4.2.1.8 Sensor coordinate system for MTx™ sensors (Mti and Mtx User Manual (2008), p. 8, Fig. 1).

4.2.2 Electromyography

Prior to electrode placement, the skin was cleansed with alcohol swabs and shaved in order to maximize sensor adherence to the skin as well as reduce skin impedence (Zipp, 1982). Following skin preparation, integral electrodes with a fixed interelectrode spacing of 2cm and an input impedence of greater than $10^{15}\Omega$ (Sensor SX230, Biometrics LTD, Cwmfelinfach, Gwent, UK) were placed bilaterally on SCM, upper trapezius and upper erector spinae muscles. For SCM muscles, electrodes were placed at 1/3 the length of the muscle from the mastoid process to the suprasternal notch (Cram, 2011; Falla et al., 2002; Sommerich et al., 2000; Zipp, 1982). For the trapezius muscles, the electrodes were placed bilaterally midway between the acromion process and spinous process of vertebra C7 (Cram, 2011; Zipp, 1982). For the cervical erector spinae muscles, electrodes were placed bilaterally at the C3 level (Shudlt & Harms-Ringdahl, 1988). The electrodes were adhered to the skin with die cut, medical grade, double-sided adhesive tape, which ensures high quality signals. A ground electrode was placed on the left olecranon process. EMG was sampled at 1000 Hz using a portable Biometrics DataLOG© system. A 4th order, dual-pass band-pass Butterworth filter was applied to the data in Microsoft Excel 2010 (Microsoft Corporation, California, USA).

4.3 Calibration

4.3.1 Kinematics

In order to calibrate the XSens® system prior to application, the MTx™ sensors were arranged on a in a straight line on a flat surface in the order they were placed on the body (i.e., forehead, sternum, C4, C7 and T4 respectively) and a 10 second sample was taken. Five minute quiet trials with each model lying both prone and supine on a flat surface were sampled simultaneously with Biometrics to provide a baseline measure (Drake & Callaghan, 2006). Reference postures consisting of 45^0 of neck flexion, 45^0 of bilateral bend to the left and right, 45^0 of bilateral rotation to the left and right, a combined 45^0 of lateral rotation and 45^0 side flexion to the left and right, as well as a shoulder shrug and a trial where the model spoke, laughed and coughed were also taken. These reference postures were taken as a contextual comparison for the kinematic data that was collected, specifically comparing the motion in these trials to the motion of interest (i.e., movement of the head into neutral position).

4.3.2 Electromyography

Prior to data collection, the Biometrics system was also calibrated and a series of maximum (100%) voluntary isometric contractions (MVCs) were taken for normalization purposes. The 5 minute supine and prone quiet trials were visually examined for the contamination of heart muscle electrical activity as those muscles in the trunk and neck in close proximity to the heart have shown to exhibit electrocardiographic artifact (Drake & Callaghan, 2004; Netto & Burnett, 2006). Three, ten second MVC trials for bilateral SCM muscles, bilateral trapezius muscles and bilateral erector spinae muscles were performed with two minutes of rest between each trial to minimize the effect of fatigue (Netto & Burnett, 2006). For each SCM muscle, the researcher provided manual resistance to the forehead as each model forward flexed and laterally rotated the neck. To collect MVCs for the trapezius muscles, the subject performed shoulder elevation against resistance provided by the researcher. For the upper erector spinae muscles, manual resistance was provided against the occiput as the models extended the necks (Shudlt & Harms-Ringdahl, 1988). The EMG trial data was normalized to the peak of the MVC values.

4.4 Data Collection Procedures

4.4.1 Pre-collection

Instrumentation and calibration of XSens® (see Sections 4.2.1, p.32 and 4.3.1, p.35) were completed first, followed by instrumentation and calibration of electromyography (see Sections 4.2.2, p.34 and 4.3.2, p.36).

4.4.2 Trials

The log roll is the most commonly used technique for prone patient transfer onto a spine board. Therefore, for the purpose of this study one Certified Athletic Therapist provided manual head-neck stabilization. During this time, three Athletic Therapists assisted in rolling the torso (at shoulder), upper and lower extremities (at hip and legs) onto a long spine board with the placement controlled by a fourth therapist. This is the typical assignment of personnel. Thirteen Certified Athletic Therapists each acted as manual stabilizers, while the other therapists involved in the roll did not change positions for the duration of that particular trial.

MTx™ sensors were placed on the models at the level of the C4 vertebra, the C7 vertebra, the T4 vertebra and the sternum to capture 3D head, and mid- low neck motion relative to the trunk as described in Section 4.2.1 (p.35). EMG of the trapezius muscle, the upper erector spinae muscles, and SCM muscles were collected bilaterally throughout the trials in order to measure muscle activation, as well as maximal efforts and responses in control postures to enable normalization.

The model's start position was held constant and was prone with the neck laterally rotated to the left. Each Certified Athletic Therapist was instructed randomly to correct neck alignment either without any specific instruction (CHOICE), DURING the log roll or AFTER the log roll for each model. Each condition was repeated three times for a total of 18 rolls. Neck motion and muscle activation was recorded throughout and the data compared for the 18 trials (3 techniques, 2 models). For this thesis, only the prone data were analyzed.

4.5 Data Processing

4.5.1 Kinematics

Angles from the level of C4 to C7 and from the level of C7 to T4 were collected and the signals were sent wirelessly through a Bluetooth connection from the XBus system to the computer. Using XAnalyzer™ software, the data was displayed in three planes (frontal, transverse and sagittal) in real time. This information was then exported and processed in Microsoft Excel 2010 (Microsoft Corporation, California, USA). The MTx™ sensors have a built in Kalman filter which provides an output of estimated drift-free 3D orientation (*MTi and MTx User Manual*, 2008). It takes the gravitational measurements from the 3D accelerometers as well as the magnetometer data (Earth's magnetic north) to offset any error caused by the angular velocity calculations from the gyroscope (*MTi and MTx User Manual*, 2008).

4.5.2 Electromyography

EMG data were collected with an eight channel Biometrics© portable data logger (Biometrics Ltd, Gwent, UK). The signal was sent wirelessly via Bluetooth to the laptop and was viewed in the Biometrics DataLog© program. It was then exported in ASCII format into Microsoft Excel 2010 (Microsoft Corporation, California, USA). The raw data were then converted from ASCII values to muscle signal values (mV) using the formula (x/4000)*3mV, where x is the ASCII value, 4000 represents the ASCII scale (*DataLOG© Operating Manual*, 2004). The EMG data were high pass filtered with a dual-pass, 4^{th} order Butterworth filter with a cutoff frequency of 30Hz (Drake & Callaghan, 2006) to remove heart rate contamination. The signals were then full wave rectified and low pass filtered with a dual-pass, 4^{th} order Butterworth filter with a cutoff frequency of 1Hz, determined by residual analysis (Winter, 2005). From the MVC trials, the maximum amplitude was used to normalize the data from the subsequent trials so the EMG signals were expressed as a percentage of MVC (%MVC).

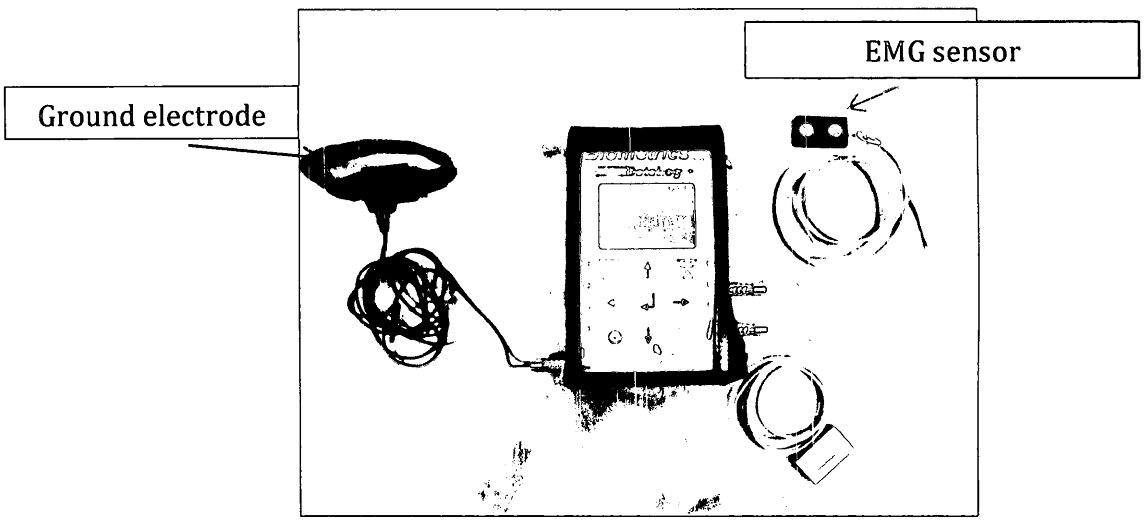

Figure 4.5.2.9 Photograph of the Biometrics Ltd. Datalog© system, including EMG sensors and ground electrode.

4.5.3 Statistical Analyses

4.5.3.1 Kinematics

After the data were processed, statistical analysis was performed in SAS 9.3 (SAS Institute Inc., Cary, NC, USA). A three-way analysis of variance (ANOVA) was run comparing the alignment correction methods (task), the two models (body size) and the amount of motion each Certified Athletic Therapist generated on the models (subject). The level of significance was set at $p=0.050$. Averages of three trials across a condition (DURING, AFTER and CHOICE) were calculated and compared for the following: range between the start and end angles (to estimate the minimum amount of required motion to correct realignment), range between the maximum and minimum angles (to determine the greatest amount of motion during the trial), and the amount of motion beyond the minimal required amount to correct alignment. Tukey's post hoc tests were performed to further analyze significant findings.

4.5.3.2 Electromyography

Statistical analyses for the EMG data were also performed using SAS 9.3 (SAS Institute Inc., Cary, NC, USA). A four-way ANOVA was run to compare muscle activity elicited due to the performance of the Certified Athletic Therapists (subject), the condition (task), the size of the model (body size) and the outputs from sensors on the left and right side of the body (side). The level of significance was set at $p=0.050$. Averages of the peak values were compared in order to monitor muscle activation across conditions. Tukey's post hoc tests were performed when significance was found in order to identify differences among combinations of variables.

5. Results

5.1 General Results

The average overall length of the trials was 24.66s (SD=2.55) for the males and 23.77s (SD=1.34) for the females. When comparing task, AFTER trials were 23.94s (SD=5.15) in length on average compared to 21.94s (4.19) for DURING trials and 24.25s (SD=9.17) for CHOICE. The average start angle across planes and sensors for females was -13.0° (SD=47.8) and for males was -11.64° (SD=37.0).

5.2 Specific Results

5.2.1 Kinematics

A three-way ANOVA was run comparing subject, task and body size of the model for theoretical minimal required motion (TMRM), additional motion beyond TMRM and range (between the maximum and minimum angles achieved). Average additional motion, TMRM, range and the maximum values achieved across tasks are shown in Table 1 (female model; f) and Table 2 (male model; m). Currently a gold standard indicating the ideal path of motion of the head during a log roll does not exist. Therefore, the theoretical minimal required motion was calculated by taking the difference between the start angle and end angle values. This was then compared between subjects, models and across tasks. A significant three-way interaction was not found for the comparison subject*task*body size ($F_{10, 600}$=0.870, p=0.564), nor a two-way interaction between task*body size ($F_{2, 600}$=0.100, p=0.903) or subject*task ($F_{20, 600}$=1.070, p=0.372). There was however a significant interaction between subject*body size ($F_{5, 600}$=11.600, 0.050> p <0.001). Out of the 11 subjects, three consistently had a higher TMRM average (63.7°, SD=32.3) than a group of seven subjects (36.8°, SD=27.4) and one subject was considerably lower than this group across body size (19.9°, SD=20.5).

When body size was addressed in the post hoc analysis, it was discovered that the female model had more significant interactions than the male. The average overall TMRM for the female model, 44.8° (SD=28.8), was higher than the male model average, TMRM of 37.4° (SD=34.3). Finally, it was found that there was no significant main effect of task ($F_{2, 600}=0.640$, $p=0.639$). Therefore, the least amount of motion required to get from the start position to the end position was not influenced by the timing of neck realignment. The average TMRM for AFTER was 42.9° (SD=31.1), for DURING was 40.5° (SD=29.8), and was 43.2° (SD=32.1) for CHOICE.

Once TMRM was calculated, any additional motion beyond this was analyzed. A three-way interaction was found for additional motion ($F_{10, 600}=2.750$, $0.0491> p <0.0048$) and for range ($F_{10, 600}=2.15$, $0.048>p<0.003$). Post hoc analysis revealed that the additional motion for one out of the eleven subjects had, on average, more motion (42.4°, SD=43.2) than the rest of the group (23.3°, SD=16.6) with the exception of the CHOICE, male and DURING, female trials . This subject was also found to have higher TMRM on average (63.6°, SD=36.8). Two other subjects had, on average, less motion than the rest of the group across all trials and both models (10.7°, SD=2.96). Using the model's CHOICE trial as the gold standard for additional motion, the percent difference was calculated. It was found that the percent difference was not significant ($F_{1, 22}=0.470$, $p=0.500$) across trials indicating that the difference in performance between subjects was not influenced by task.

When analyzing the interaction across tasks for additional motion, it was found that more motion on average was elicited during the AFTER and DURING trials as opposed to the

CHOICE trials across subjects and body size. The average amount of motion for AFTER trials was 25.7° (SD=23.9), for DURING trials was 23.5° (SD=22.8) and for CHOICE was 20.9° (SD=16.5). Of the subjects who had the most amount of significant interactions, two of the eleven caused more motion, on average for AFTER trials (52.1°, SD=45.8). Yet only one of these particular subjects had higher average TMRM as well (62.5°, SD=29.4) and the start position of the models was the same. Of those interactions where the average amount of additional motion was lower than the rest of the group of subjects, the averages were equally distributed across tasks (11.93°, SD=9.53 AFTER, 10.83°, SD=6.71 DURING and 9.4°, SD=5.23 CHOICE).

When looking at the body size factor in the three-way interaction, the subjects caused an almost equal amount of motion for the female and male model across task. The average amount of additional motion overall created in the female model trials was 21.8° (SD=19.0) and 27.2° (SD=25.0) for the male model trials. This suggests that the size of the model does not have a major influence on the differences between subjects across tasks.

There was also a three-way interaction ($F_{10,597}$=2.15, 0.048>p<0.003) when using comparing the ranges between the maximum and minimum values as the dependent variable across tasks and body sizes of the models. During the post hoc analysis, it was found that two subjects out of the eleven consistently had a higher range than the rest of the group, while one subject was consistently lower, both across tasks and body sizes of the models. These subjects are the same ones who had higher and lower TMRM on average as well as higher and lower additional motion on average. The average ranges of motion were as follows: 90.1° (SD=30.8) for the two subjects with the largest range, 38.8° (SD=17.7) for the subject with the smallest range and 61.8° (SD=29.5) for the rest of the group (n=8). When using CHOICE as the gold standard to calculate the percent difference between AFTER and DURING for range, no significance was found ($F_{1, 22}$=0.420, p=0.526). Similar to additional motion, this finding suggests that the differences between how the subjects performed head realignment were consistent regardless of the task.

Using task as the main focus in the interaction, with differences across subjects and body size, the AFTER trials had a slightly higher overall averages than CHOICE and DURING. The overall average for AFTER range was 69.0° (SD=33.1), for DURING range was 60.6° (SD=30.2) and for CHOICE range was 62.0° (SD=29.3). For three subjects who consistently had significant interactions, AFTER range (102.5°, SD=30.1) was higher than DURING and CHOICE range for the rest of the group (59.9°, SD=28.5).

In analyzing body size in the three-way interaction, when the female model was moved, the CATs generated an average range of 66.2° (SD=29.6), while on average, when the male model was moved, range of 59.2° (SD=33.2) was generated. Of those subjects who had consistently significant interactions, the female model range (92.3°, SD=32.0) was lower than male model range (98.5°, SD=34.6). The values are higher than the averages for the rest of the subjects which were 61.3° (SD=28.1) for the female model and 51.1° (SD=26.0) for the male model.

The maximum angles achieved, shown in Tables 5.2.1.4 and 5.2.1.8, occurred on average at 53.5% (SD=15.84) of the length of the AFTER trials, at 53.3% (SD=20.5) of the length of the DURING trials and at 47.9% (SD=21.69) of the length of the CHOICE trials for the male and at 51.1% (SD=19.4), 50.9%, (SD=21.2) and 51.1% (SD=20.5) respectively for the female model. The minimum values occurred at 69.6% (SD=6.67) for male AFTER, 73.2% (SD=5.9) for male DURING and 65.6% (SD=10.5) for male CHOICE. Whereas, the female minimum values occurred at 52.0% (SD=7.5) for the AFTER trials, 54.6% (SD=8.5) for DURING trials, and 51.7% (SD=10.2) for CHOICE trials. Figures 5.2.1.10, 5.2.1.11 and 5.2.1.12 show sample traces of kinematic motion for the female model and the forehead sensor for the different timing conditions. Although some variability between the starting positions was evident, it was not significant.

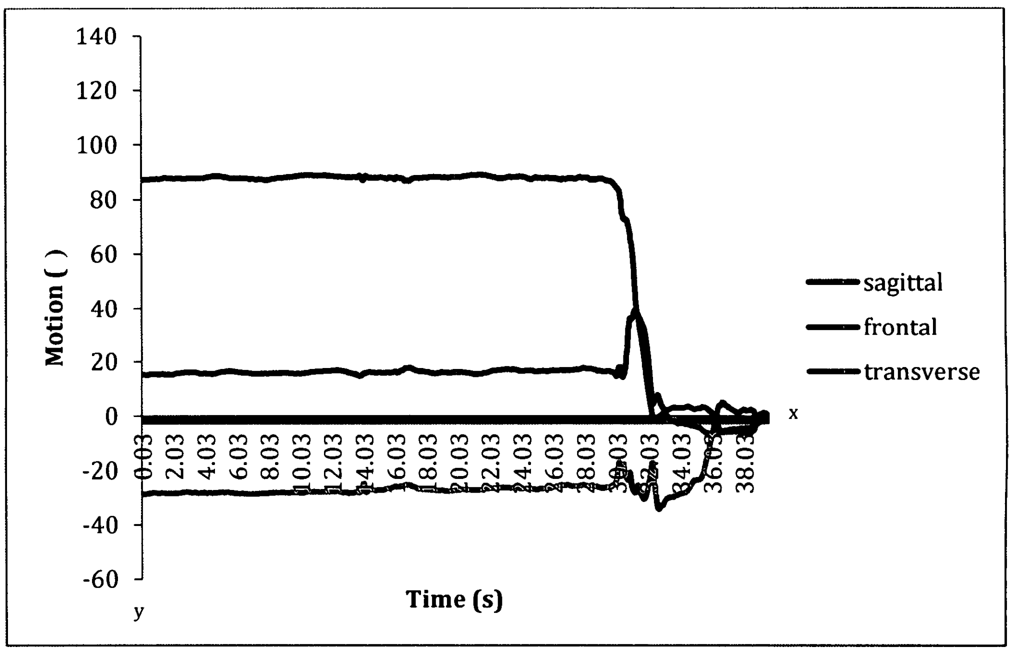

Figure 5.2.1.10 Example of kinematic motion (°) of the forehead over the trial length (s) for the AFTER condition, female model.

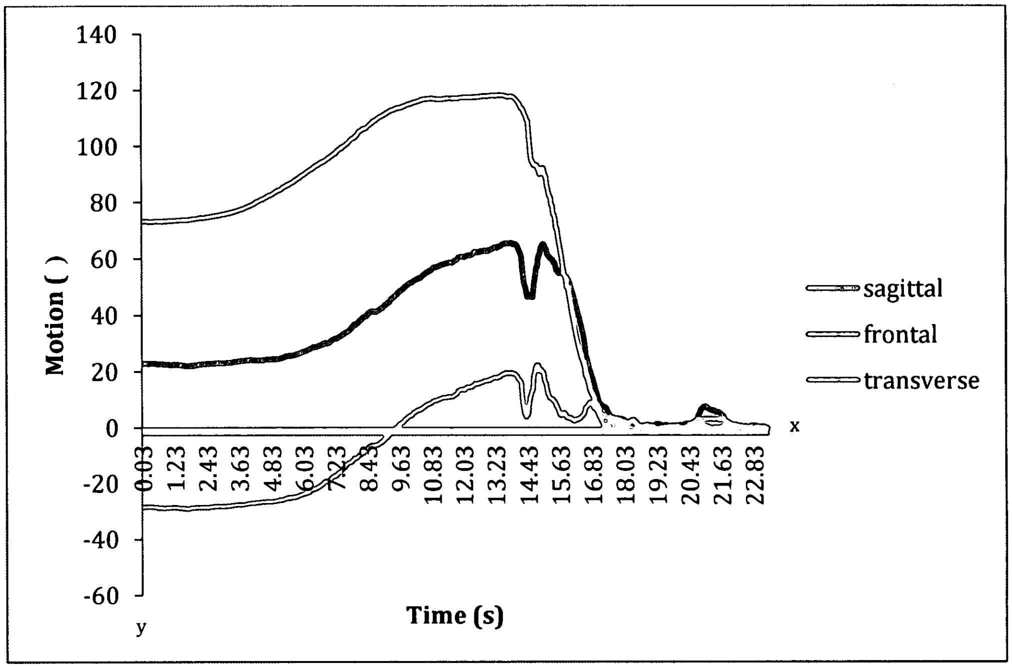

Figure 5.2.1.11 Example of kinematic motion (°) of the forehead over the trial length (s) for the DURING condition, female model.

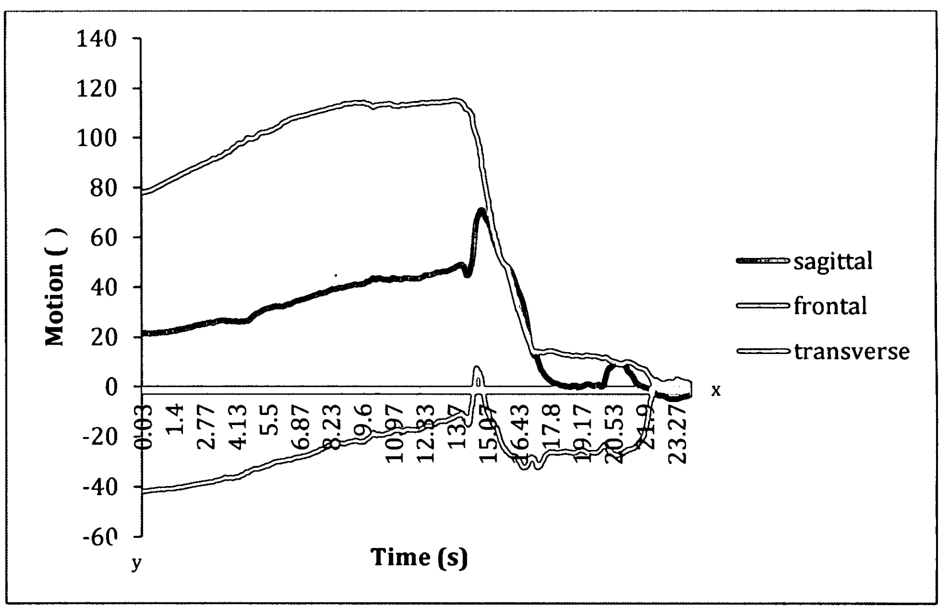

Figure 5.2.1.12 Example of kinematic motion (°) of the forehead over the trial length (s) for the CHOICE condition, female model.

This example demonstrates the trend of timing of motion and shows that two separate conditions were actually elicited from the Certified Athletic Therapists. In the three graphs above, one can clearly see the different conditions (correct AFTER and correct DURING). The realignment phase happens earlier in the DURING trial (at roughly 52% versus 62% for the AFTER trial).

Table 5.2.1.1 Female model kinematic averages additional motion shown in degrees

	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	18.3, SD=12.8	21.6 SD=16.0	16.0, SD=9.16	19.0, SD=25.2	15.6, SD=11.5	9.8, SD=5.4	17.9, SD=14.7	13.2, SD=10.4	9.9, SD=7.4
C4	31.6, SD=16.3	18.7, SD=17.2	26.4, SD=22.3	29.2, SD=20.6	14.9, SD=14.7	17.8, SD=16.3	23.3, SD=12.3	18.2, SD=14.6	16.3, SD=14.3
C7	35.5, SD=22.2	13.6, SD=9.4	22.5, SD=17.6	32.8, SD=16.8	14.5, SD=12.9	19.4, SD=15.8	26.9, SD=14.8	14.5, SD=14.4	20.8, SD=17.2
T4	33.5, SD=13.9	16.4, SD=9.0	21.4, SD=16.6	32.6, SD=15.1	14.3, SD=12.2	19.3, SD=13.8	31.3, SD=14.5	17.6, SD=22.3	24.7, SD=25.9

Table 5.2.1.2 Female model kinematic average TMRM shown in degrees

	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	19.7, SD=10.1	23.8, SD=26.8	47.5, SD=21.2	20.23, SD=10.7	21.1, SD=22.7	47.8, SD=20.4	23.6, SD=13.3	31.5, SD=30.8	48.9, SD=18.0
C4	23.5, SD=13.1	62.0, SD=29.1	47.6, SD=22.4	25.7, SD=14.4	53.4, SD=27.8	45.5, SD=26.7	24.7, SD=17.1	61.8, SD=19.5	50.8, SD=24.8
C7	29.4, SD=16.3	68.2, SD=23.9	62.1, SD=29.6	30.9, SD=21.5	57.7, SD=27.3	58.8, SD=27.2	41.1, SD=25.5	70.3, SD=21.7	64.7, SD=29.7
T4	24.2, SD=21.0	59.0, SD=33.7	73.3, SD=27.6	24.4, SD=24.7	56.3, SD=31.3	67.6, SD=25.8	31.2, SD=30.2	66.8, SD=30.81	76.93, SD=21.6

Table 5.2.1.3 Female model kinematic range averages shown in degrees

	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	37.9, SD=10.4	45.4, SD=27.5	60.8, SD=22.1	40.7, SD=24.6	39.1, SD=35.3	57.5, SD=31.2	40.1, SD=13.7	42.3, SD=31.2	58.9, SD=20.5
C4	55.1, SD=20.9	80.6, SD=30.9	71.9, SD=30.9	54.7, SD=25.4	70.3, SD=26.3	64.1, SD=23.7	48.2, SD=18.1	70.8, SD=26.3	60.1 SD=28.3
C7	65.0, SD=25.1	77.0, SD=18.6	84.6, SD=32.0	63.9, SD=26.0	75.6, SD=25.7	77.3, SD=32.0	66.0, SD=29.7	72.8, SD=21.8	77.0, SD=32.6
T4	57.7, SD=26.1	75.4, SD=29.8	94.8, SD=32.0	56.6, SD=25.5	73.9, SD=32.6	85.9, SD=26.1	56.5, SD=27.1	66.0, SD=34.1	85.5, SD=29.8

Table 5.2.1.4 Female model kinematic maximum angle averages shown in degrees

	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	28.1, SD=26.0	23.4, SD=14.3	40.2, SD=33.8	28.6, SD=24.6	20.1, SD=20.3	37.6, SD=30.5	30.0, SD=26.0	23.2, SD=15.3	38.0, SD=31.6
C4	9.0, SD=17.8	48.7, SD=17.8	52.2, SD=24.3	13.9, SD=24.6	44.0, SD=22.8	44.7, SD=24.0	9.2, SD=20.7	43.2, SD=23.3	41.4, SD=12.7
C7	4.2, SD=8.7	29.7, SD=14.9	46.8, SD=21.4	5.8, SD=11.3	32.4, SD=20.4	44.3, SD=27.0	6.5, SD=12.7	30.0, SD=19.2	44.3, SD=27.2
T4	7.3, SD=12.4	30.6, SD=12.4	59.8, SD=16.3	7.8, SD=14.7	29.3, SD=17.6	52.8, SD=25.2	9.9, SD=15.3	27.5, SD=13.3	59.0, SD=19.1

Table 5.2.1.5 Male model kinematic additional motion averages shown in degrees

	ADDITIONAL MOTION (°)								
	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	28.2, SD=7.1	32.1, SD=22.0	15.0, SD=8.5	26.7, SD=7.8	18.0, SD=14.5	22.3, SD=7.3	23.5, SD=7.6	19.0, SD=16.4	14.1, SD=8.1
C4	34.3, SD=13.9	22.4, SD=9.7	20.8, SD=14.6	31.9, SD=9.1	19.3, SD=7.7	17.6, SD=11.5	28.3, SD=10.3	13.8, SD=6.1	13.8, SD=8.4
C7	28.6, SD=5.1	19.1, SD=10.1	26.4, SD=12.1	31.1, SD=7.1	21.4, SD=10.4	24.3, SD=14.8	24.3, SD=8.8	18.4, SD=11.5	19.8, SD=11.8
T4	37.72 SD=13.8	20.2, SD=5.5	26.9, SD=12.2	33.5, SD=7.5	21.8, SD=7.8	26.7, SD=14.0	35.5, SD=9.0	21.6, SD=12.0	20.1, SD=9.7

Table 5.2.1.6 Male model kinematic TMRM averages shown in degrees

	TMRM (°)								
	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	45.2, SD=48.9	13.6, SD=7.5	63.5, SD=31.2	39.0, SD=38.2	17.5, SD=10.1	58.1, SD=35.1	37.7, SD=50.3	14.8, SD=9.8	53.0, SD=19.3
C4	28.2, SD=32.8	36.0, SD=35.2	47.5, SD=39.3	24.7, SD=33.3	32.1, SD=29.7	44.8, SD=36.2	18.6, SD=18.6	38.5, SD=36.0	45.8, SD=41.23
C7	30.8, SD=34.4	46.4, SD=35.6	50.8, SD=38.0	28.6, SD=33.3	44.1, SD=30.9	49.5, SD=40.6	30.3, SD=35.0	44.0, SD=38.6	46.4, SD=35.9
T4	19.8, SD=27.3	44.1, SD=26.4	54.3, SD=44.2	21.0, SD=33.7	38.2, SD=21.6	52.5, SD=46.4	17.2, SD=20.7	41.0, SD=29.7	40.5, SD=37.7

Table 5.2.1.7 Male kinematic range averages shown in degrees

	RANGE (°)								
	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	19.8, SD=27.3	44.1, SD=26.4	54.3, SD=44.2	21.0, SD=33.7	38.2, SD=21.6	52.5, SD=46.4	17.2, SD=20.7	41.0, SD=29.7	40.5, SD=37.7
C4	62.5, SD=46.7	58.5, SD=32.5	68.3, SD=41.7	56.6, SD=41.3	51.6, SD=30.6	62.5, SD=43.0	46.9, SD=25.2	52.3, SD=31.5	59.6, SD=39.3
C7	59.5, SD=38.1	65.5, SD=31.0	77.2, SD=44.7	59.7, SD=39.5	65.6, SD=36.3	73.8, SD=49.2	54.6, SD=31.8	62.4, SD=35.1	66.3, SD=38.8
T4	57.5, SD=29.4	64.38 SD=23.8	81.2, SD=49.0	54.4, SD=34.0	60.1, SD=27.5	79.2, SD=51.9	46.5, SD=19.3	57.9, SD=30.2	67.4, SD=39.2

Table 5.2.1.8 Male model kinematic maximum angle averages shown in degrees

	MAXIMUM ANGLE (°)								
	AFTER			DURING			CHOICE		
	X	Y	Z	X	Y	Z	X	Y	Z
FH	40.2, SD=22.8	25.8, SD=28.6	38.2, SD=38.6	50.2, SD=40.3	24.3, SD=20.2	41.0, SD=33.8	55.0, SD=62.7	24.1, SD=19.2	33.43, SD=36.8
C4	5.4, SD=11.5	35.1, SD=23.6	49.9, SD=32.3	0.6, SD=0.8	28.4, SD=24.2	39.8, 30.0	0.5, SD=1.2	30.2,SD=23.6	40.1, SD=27.9
C7	-36.2, SD=37.6	16.2 ,SD=24.0	44.5, SD=32.4	19.9, SD=48.7	19.4, 27.6	38.3, SD=29.8	-37.6, SD=10.0	18.1, SD=23.4	36.2, SD=30.8
T4	5.6, SD=8.2	18.0, SD=22.1	0.8, SD=30.4	2.9, SD=5.1	17.5, SD=25.1	47.8, SD=29.6	1.1, SD=1.6	16.8, SD=24.6	45.4, SD=29.0

5.2.2 Electromyography

A four-way ANOVA was run comparing the peak amplitudes of EMG (%MVC) across subjects, tasks, body size of the models and between the outputs of the sensors located on the left versus the right hand side of the body. Tables 5.2.2.9 to 5.2.2.11 show the average peak muscle activation (%MVC) by side as well as the average timing of peak muscle activation (% of the total trial length) for female and male models across tasks. There was no significant four-way interaction found between subject*task*body size*side ($F_{15, 262}$=0.280, *p*=0.996). There were no significant three-way interactions found for subject*task*body size ($F_{10, 262}$=0.560, *p*=0.847), task*body size*side ($F_{3, 262}$=0.110, *p*=0.953), nor for subject*task*side ($F_{20, 262}$=0.180, *p*=0.9525). There were not significant two-way interactions between task*side ($F_{2, 262}$=0.050, *p*=0.948), task*body size ($F_{2, 262}$=0.028, *p*=0.758), subject*task ($F_{20, 262}$=0.054, *p*=0.942) or subject*side ($F_{10, 262}$=0.730, *p*=0.699). There was however a significant two-way interaction between subject and body size ($F_{5, 262}$=4.070, 0.043>p<0.032). Post hoc analysis revealed that one subject elicited significantly more muscle activity as a %MVC for the female model (45.7 %MVC, SD=30.49) than the rest of the group (11.08 %MVC, SD=15.97). Please note, this difference is suspected to be due to the subject not the MVC protocol of the model because two additional subjects were collected in the same day and did not elicit the same trend (25.4%MVC, SD=26.3). The female model had higher peak muscle activation (%MVC) overall (17.01 %MVC, SD=20.35) compared to the male subject (3.78 %MVC, SD=4.62). This suggests that the male model did a better job at staying completely relaxed during the trials. Figure 5.2.2.13 shows the distribution of peak EMG amplitude across sensors for both models. The greatest maximum amplitude was for RES female (32.61 %MVC, SD=29.02 %MVC) and the smallest maximum amplitude was for sensor RUT, male (0.64 %MVC, SD=1.84 %MVC).

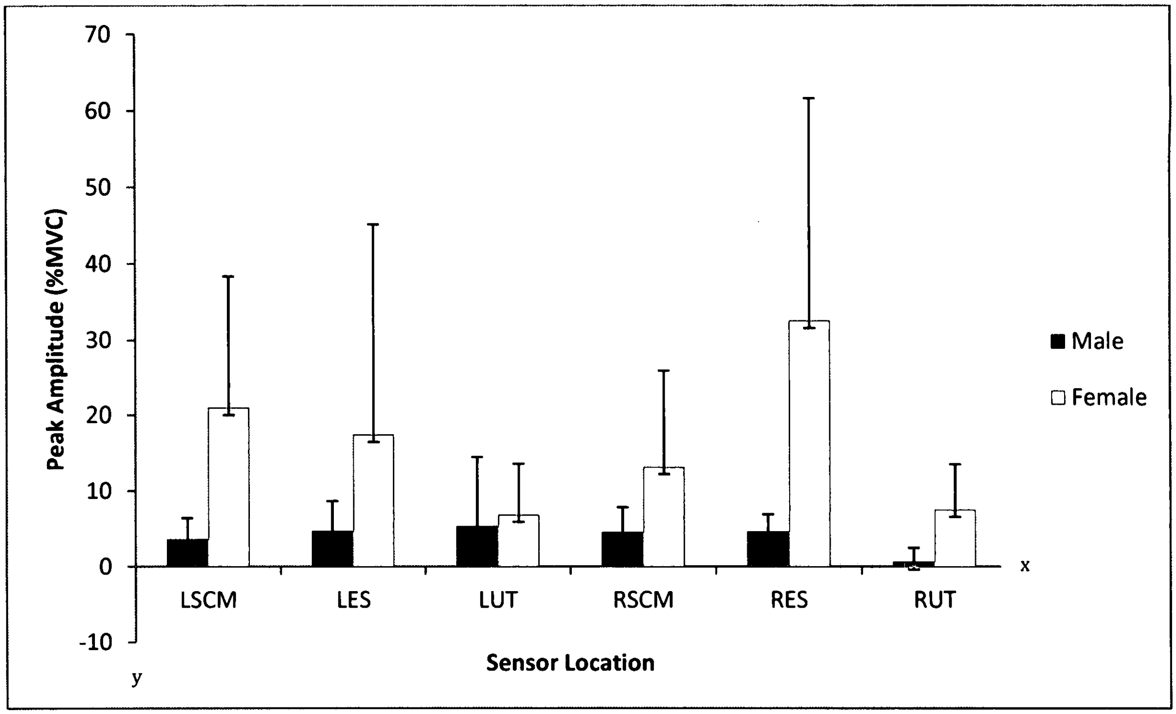

Figure 5.2.2.13 Comparison of peak EMG amplitude (%MVC) across left and right sternocleidomastoid, erector spinae and upper trapezius sensors of both male and female model.

Peak muscle activation was not significant by task ($F_{2, 262}$=0.940, p=0.392) or when comparing sides of the body ($F_{1, 262}$=0.560, p=0.456). The average peak muscle activation was 20.17 %MVC (SD=18.9) for AFTER, 10.96 %MVC (SD=17.3) for DURING, and 11.78 %MVC (SD=16.92) for CHOICE. Muscle activation was almost the same for left (11.60 %MVC, SD=16.15) and right sides (12.88%MVC, SD=18.91) of the body. The average EMG activation across trials were 2.73%MVC (SD=2.3.4) for AFTER, 2.16 %MVC (SD=1.42) for DURING and 2.03 %MVC (SD=1.38) for CHOICE.

As a point of interest, overall male peak muscle activation occurred at 52% (SD=15.4) of the trial length versus the female whose peak muscle activation occurred at 23.4% (SD=4.8).

The average Figure 5.2.2.14 shows the overall average of the time to peak amplitude (% of total trial length) for male (left and right) and female (left and right).

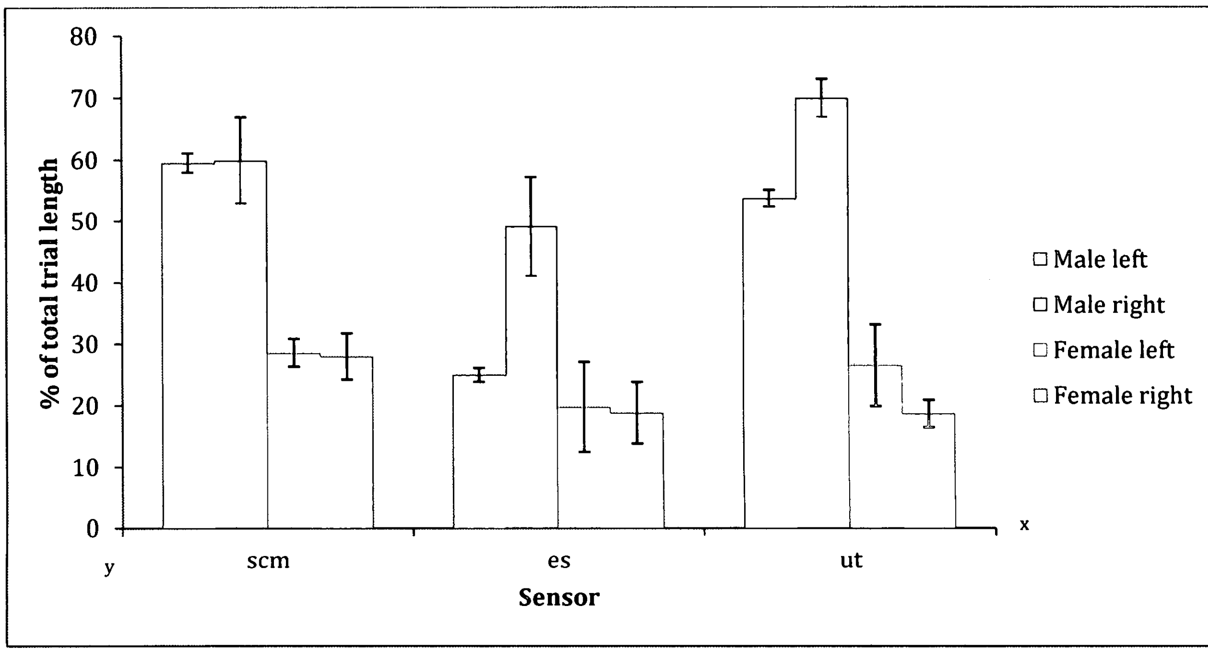

Figure 5.2.2.14 Time to peak amplitude between male and female models by side (% of total trial length).

Table 5.2.2.9 Average peak amplitude (%MVC) and time to peak amplitude (%) for left and right SCM

	TASK	AFTER	DURING	CHOICE	AFTER	DURING	CHOICE
	SENSOR LOCATION	LSCM	LSCM	LSCM	RSCM	RSCM	RSCM
FEMALE	MAX VALUE	26.5, SD=18.6	17.3, SD=14.0	23.3, SD=19.3	18.4, SD=16.8	8.9, SD=3.8	15.7, SD=14.4
	MAX TIME %	66.8, SD=17.7	63.0, SD=13.6	52.0, SD=27.8	54.6, SD=17.4	50.9, SD=14.7	59.0, SD=24.3
MALE	MAX VALUE	2.9, SD=1.6	5.1, SD=4.1	2.8, SD=1.6	3.8, SD=1.4	5.1, SD=4.9	3.1, SD=0.9
	MAX TIME %	60.6, SD=18.7	63.2, SD=20.9	54.7, SD=21.7	66.2, SD=10.7	57.4, SD=23.3	56.1, SD=11.8

Table 5.2..2.10 Average peak amplitude (%MVC) and time to peak amplitude (%) for left and right ES

	TASK	AFTER	DURING	CHOICE	AFTER	DURING	CHOICE
	SENSOR LOCATION	LES	LES	LES	RES	RES	RES
FEMALE	MAX VALUE	10.6, SD=23.9	12.8, SD=27.2	2.2, SD=3.2	34.2, SD=28.7	33.6, SD=30.5	32.5, SD=29.2
	MAX TIME %	24.3, SD=31.1	14.6, SD=18.9	21.7, SD=29.4	58.0, SD=20.3	44.3, SD=19.3	48.2, SD=16.1
MALE	MAX VALUE	0.7, SD=1.8	1.6, SD=3.8	0.82, SD=1.91	0.1, SD=1.3	5.3, SD=0.3	4.9, SD=1.9
	MAX TIME %	22.5, SD=38.7	25.9, SD=40.1	26.4, SD=41.0	39.1, SD=28.5	63.6, SD=12.5	44.5, SD=18.3

Table 5.2.2.11 Average peak amplitude (%MVC) and time to peak amplitude (%) for left and right UT

	TASK	AFTER	DURING	CHOICE	AFTER	DURING	CHOICE
	SENSOR LOCATION	LUT	LUT	LUT	RUT	RUT	RUT
FEMALE	MAX VALUE	7.5, SD=7.2	3.7, SD=4.1	5.9, SD=7.7	7.2, SD=7.2	4.9, SD=4.4	6.8, SD=7.4
	MAX TIME %	48.2, SD=30.1	42.4, SD=26.9	51.7, SD=31.3	41.8, SD=30.6	35.5, SD=25.3	39.5, SD=24.9
MALE	MAX VALUE	4.5, SD=9.0	4.5, SD=9.2	4.13, SD=9.19	0.1, SD=0.1	0.2, SD=0.1	0.1, SD=0.1
	MAX TIME %	58.5, SD=30.7	53.8, SD=28.1	48.6, SD=29.2	61.3, SD=13.4	71.6, SD=10.8	76.9, SD=7.3

5.3 Reference Postures

The reference postures were also processed and compared. The average length of time to complete the 9 reference postures was 11.77s (SD=1.38) for females and 9.53 s. (SD=1.14 s). The reference postures, maximum angle achieved and the greatest difference from the start angle to the maximum angle are seen below in Table 5.3.12.

Table 5.3.12 Summary of average maximum angles achieved during reference postures for the female model (A) and the male model (B).

(A)

Female Reference Posture	Neck Movement	Maximum angle, sensor, plane	Range (start value to maximum angle)
1	Forward flexion 45°	32.8° (SD=47.9), FH, x	35.7°, FH, x
2	Lateral rotation left	48.76° (SD=16.11), FH, z	48.6°, FH, z
3	Lateral rotation right	26.17° (SD=19.45), C4, z	26.7°, T4, x
4	Left side bend	25.78° (SD=8.08), FH, z	35.3°, C7, x
5	Right side bend	31.5°(SD=7.59), C4, z	29.7°, C4, x
6	Combination lateral rotation, side bend left	20.85° (SD=7.06), C4, z	13.5°, C4, y
7	Combination lateral rotation, side bend right	39.26° (SD=18.4), C7, x	20.6°, FH, z
8	Shoulder shrug	29.44° (SD=16.25), C7, x	10.8°, FH, x
9	Cough, swallow, speech	30.56° (SD=17.01), C4, x	9.78°, C7, y

(B)

Male Reference Posture	Neck Movement	Maximum angle, sensor, plane	Range (start value to maximum angle), sensor, plane
1	Forward flexion 45°	-36.25°(SD=25.48°), C7, x	33.6°, FH, x
2	Lateral rotation left	-49.87°(SD=11.07°), C7, x	41.4°, FH, z
3	Lateral rotation right	-55.72° (SD=1.57°), C7, x	35.6°, FH, z
4	Left side bend	-58.2° (SD=4.68°), C7, x	38.5°, FH, x
5	Right side bend	-56.37 (SD=3.54), C7, x	28.3°, FH, z
6	Combo lateral rotation, side bend left	-46.26(SD=5.24), T4, x	16.3°, FH, z
7	Combo lateral rotation, side bend right	-53.4(SD=13.73), C7, x	15.4°, C4, z
8	Shoulder shrug	-49.97(SD=8.01), C7, x	16.5°, FH, z
9	Cough, swallow, speech	-53.6(SD=0.71), C7, x	11.75, FH, z

Four EMG maximum voluntary contraction trials were completed during the pretrial sessions and the average MVCs are shown in Table 5.3.13 below.

Table 5.3.13 Maximum voluntary contractions achieved during pretrial collection (mV)

SENSOR	FEMALE	MALE
LSCM	0.696, SD=0.488	0.468, SD=0.094
RSCM	0.548, SD=0.307	0.369, SD=0.088
LES	0.320, SD=0.314	0.777, SD=0.615
RES	0.144, SD=0.056	0.245, SD=0.079
LUT	0.801, SD=0.457	1.136, SD=0.991
RUT	0.790, SD=0.483	2.371, SD=0.106

The reference postures were examined for muscle activity. The average amount of overall muscle activation ranged from 0.4%MVC (SD=0.6) for RUT, reference posture 3 to 28.7%MVC (SD=38.0) for RES, reference posture 3 for the male model and 0.2%MVC (SD=0.05) for LUT, reference posture 4 and RUT, reference posture 5 to 35.5%MVC (SD=36.4) for RES, reference posture 2 for the female.

In terms of kinematics, the average maximum angles during the reference postures are shown in Table 5.3.15. The average maximum angle was 30.64° (SD=27.92) for the female model and 25.72° (SD=30.45°) for the male model during the trials. This indicates that the motion created during a log roll is unlike simple rotation to the left and then back to neutral.

Table 5.3.14 Reference posture average maximum angles (°) for female and male model

REFERENCE POSTURE	MAXIMUM ANGLE(°), F	MAXIMUM ANGLE (°), M
1	7.45, SD=16.07	-3.51, SD=17.31
2	6.42, SD=23.89	0.34, SD=30.29
3	10.7, SD=7.16	-4.39, SD=26.62
4	11.45, SD=8.67	-3.86, SD=25.98
5	3.10, SD=17.49	-7.10, SD=26.12
6	1.42, SD=17.38	-3.59, SD=24.78
7	0.65, SD=18.68	-6.01, SD=24.87
8	2.04, SD=18.36	-7.48, SD=25.26
9	-0.77, SD=18.18	-10.07, SD=30.35

Figure 5.3.14 shows an example of kinematic motion during reference posture 2 (left lateral rotation) for the forehead sensor, female model. If we revisit Figures 5.2.1.10 through 5.2.1.12 and compare them to Figure 5.3.14, there we can confirm that there are not any similarities.

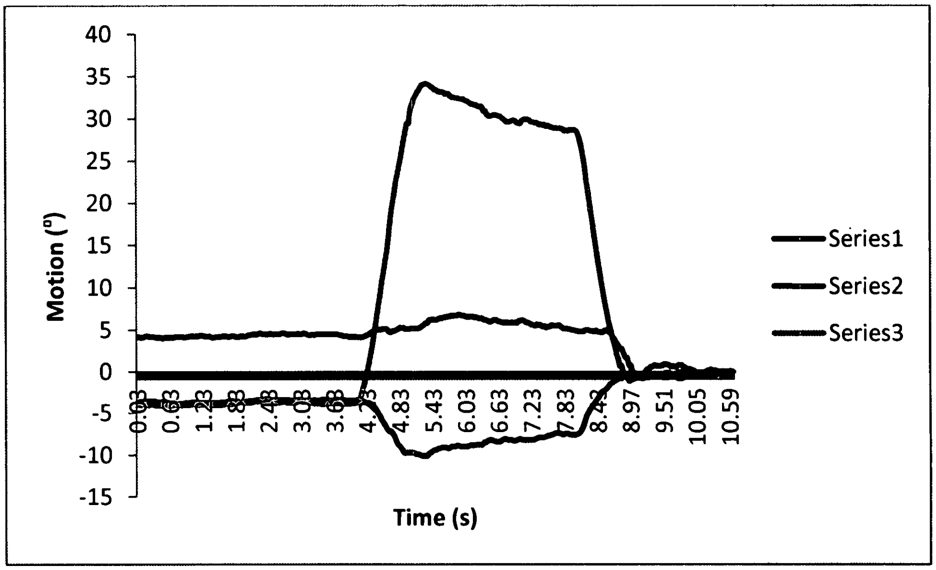

Figure 5.3.15 Example of kinematic motion (°) of the forehead sensor, female model over time (s) for reference posture 2 (left lateral rotation).

6. Discussion

This purpose of this study was to investigate the amount of cervical spine motion that takes place during the alignment of the head and neck occurring during the initial management stages of an acute cervical spine injury. Proper management of the acute SCI is necessary to reduce the risk of secondary injury to the spinal cord (Banerjee et al., 2005). It is the primary responsibility of first responders to maintain spinal stabilization until an appropriate evaluation can be performed. This study will impact all those professions providing acute emergency care of those individuals with a suspected SCI. It will help enhance knowledge of transfer techniques on those with spinal injuries as well as helping to develop proper training techniques for those involved in the primary management stages of a victim with a suspected spinal cord injury. Current research suggests the avoidance of the log roll technique altogether as it causes increased motion compared to other techniques (Conrad et al., 2012). However, there have not been any alternative transfer methods recommended and with appropriate research for the prone victim. This study was unique in that it analyzed motion at the cervical spine during prone patient transfer onto a long spine board using the log roll technique. Currently, there is a gap in the literature with regard to this specific portion of the initial management stages of a suspected SCI.

6.1 Hypotheses Revisited

The first hypothesis was that the timing of neck realignment does modify the amount of cervical spine motion and muscle activation during boarding. Specifically, alignment of the neck during the transfer will cause less motion and muscle activation as those stabilizing the head will not be able to maintain stabilization in the position found for the duration of the roll. Although a three-way interaction for subject*task*body size was found for both range and additional motion, percent differences were compared between DURING and AFTER trials and were not found to be significantly different. Peak muscle activity was also analyzed and no significant differences were found across tasks. We can therefore accept the null hypothesis that the timing of neck realignment does not play a role in the amount of cervical spine motion during the patient transfer process.

The second hypothesis was that the weight and size of the patient does not play a role in the amount of cervical spine motion and muscle activation during patient transfer onto a spine board. TMRM, additional motion and range were analyzed. Significant three-way interactions for subject*task*body size were found for both additional motion and range and a significant two-way interaction was found for TMRM between subject and body size. Upon further analysis, it was found that overall the average amount of additional motion, range and TMRM generated when female model was rolled was consistently higher than that of the male victim. In terms of EMG, the male model had a lower overall average %MVC across sensors. Therefore, the null hypothesis, that the weight and size of the victim have an influence on the amount of cervical spine motion generated during the log roll, can be accepted. Though this is an interesting finding, it should not necessarily affect the boarding process itself as the goal is still and always will be to create the least amount of motion possible. In order to determine if body size truly has an effect on the amount of motion during the log roll, data needs to be collected using more models of similar weights and sizes to the models used in this study as well as individuals whose weight and size fall between them (which are considered to be at end ranges of the spectrum).

6.2 Kinematics

Figures 5.2.1.10, 5.2.1.11 and 5.2.1.12 showed sample traces of kinematic motion at the forehead for the different timing conditions for the female model. From these examples, it is evident that there are three distinct phases that are common across tasks for both the male and female models. First, there is an initial plateau across planes as the Certified Athletic Therapist stabilizes the head preparing for the log roll. The next phase shows when the realignment is happening and finally there is a levelling off again when the head is placed in neutral. Although the timing of realignment is clearly evident in those figures, there were no significant differences in overall TMRM, range and additional motion across the three different conditions.

The major goal of this study was for the Certified Athletic Therapist to get the model's head from the start position (prone with the head laterally rotated) to the end position (supine, head in neutral position) with the least amount of movement possible. Therefore, theoretical minimal required motion, TMRM, was calculated and used as a baseline and to represent the gold standard since none currently exists. This was done by subtracting the start angle from the end angle. The average least amount of motion required to get from the start position to the end position was 42.22° (SD=31.02) across all sensors and planes. TMRM was compared across tasks as a starting point to move forward and analyze any additional motion beyond this theoretical "straight line motion". The specific values for each plane and task for the female and male models are seen in Tables 5.2.1.2 and 5.2.1.6 respectively. It was found that there was no significant difference in TMRM across tasks ($p=0.639$). The average TMRM for AFTER was 42.9° (SD=31.1), for during was 40.5° (SD=29.8), and was 43.2° (SD=32.1) for choice. This finding makes sense because the models started and ended in the same positions, regardless of the timing of realignment. Calculation of TMRM was a necessary preliminary step in order to address the hypothesis that the timing of neck realignment (DURING or AFTER the log roll) will modify the amount of cervical spine motion.

Another important preliminary finding is the range between the maximum and minimum angles achieved across trials. The average ranges elicited on the female and male model are seen in Tables 5.2.1.3 and 5.2.1.7. A significant three-way interaction was found between subject, body size of the model and range. Out of the 11 subjects, two consistently had higher ranges (90.1°, SD=30.8) than the rest of the group which had an average overall range of 38.8° (SD=17.7), with the exception of one subject who had the smallest range of motion (61.8°, SD=29.5) across body size and task. In order to focus on the differences across tasks, the CHOICE trial was used as the gold standard and the percent differences were calculated for DURING and AFTER. It was again found that there were no significant differences between the tasks. The overall averages were found to be 69.0° (SD=33.1) for AFTER, 60.6° (SD=30.2) for DURING and 62.0° (SD=29.3). These ranges between maximum and minimum are very similar. Therefore, we can conclude that task does not have an influence on the differences between the average ranges of motion. Once TMRM and range were calculated and analyzed, they were used to calculate the total amount of motion that actually occurred during the trials. In order to quantify the amount of motion that occurred during this study additional motion was calculated by taking the difference between the TMRM and range of the maximum and minimum values. When comparing this additional motion across subjects, task and model size, a significant three-way interaction was found. One subject had more motion on average than the group (42.4°, SD=43.2 compared to 23.3°, SD=16.6). Also, two subjects had less motion (10.7°, SD=2.96). This means that the Certified Athletic Therapists did not perform the exactly the same, which was to be expected because, although they all had the same amount of technical emergency care training, they had all been certified for different lengths of time (see Section 4.1, p. 31). In fact, it is interesting to note that the subject who generated the least amount of additional motion and smallest range was the least experienced therapist in the group. Also interesting, this finding is more than likely due to systematic differences rather than functional. Therefore, the focus was turned to determining whether the difference between tasks influenced the amount of motion. In order to do this, the CHOICE trial was used as the gold standard and the percent difference between the AFTER and DURING trials were calculated and compared across subjects. In doing so, it was determined that there were no differences across tasks. In looking strictly at the averages of additional motion by task, they were very similar (see Tables 1 and 4 for male and female averages). The average amount of motion for AFTER trials was 25.7° (SD=23.9), for DURING trials was 23.5° (SD=22.8) and for choice was 20.9° (SD=16.5).

Controversy exists in current practice for the timing of prone victim transfer onto a spine board. Common practice includes correcting misalignment of the neck using the method that is most comfortable for the therapist responsible for realigning the head. During this study, when given the choice to correct DURING or AFTER, 81% of the trials overall were corrected DURING the roll. Eighty-nine percent of the trials with the male model were corrected DURING, while only 77% of the female model trials were correct DURING. This suggests that the preferred method of timing, especially when there is a considerable weight difference between the models, is to correct DURING the roll. As this study has shown, there is no difference in the amount of motion that occurs based on the timing of realignment. Therefore, the head Athletic Therapist can naturally perform the log roll using the timing of his or her choice and focus on additional factors that may influence the outcome of the roll. Such factors include assessment and direction of the team of the other individuals involved in the boarding process.

The second hypothesis was that the weight and size of the victim does not play a role in the amount of cervical spine motion during the transfer process. In order to address this hypothesis, the amount of theoretical minimal required motion, range and additional motion were analyzed. With regard to TMRM, there was a significant interaction found between subject and body size. TMRM was consistently higher for three of the subjects across body size of the model (63.7°, SD=32.3) and one subject had considerably lower average TMRM (19.9°, SD=20.5) between the body sizes of the models. In addressing a specific example between the highest TMRM subject (with an average TMRM of 71.7°, SD=33.2) and the lowest (with an average TMRM of 19.86°, SD=20.46), it is interesting to note that although there was a significant difference between the average TMRM ($p<0.0001$), when additional motion was considered along with TMRM, the subject with the consistently high average TMRM was actually fairly efficient at maintaining stabilization as the additional motion constituted only 27.7% of overall motion. The subject who had a consistently low TMRM had 55% of the overall motion accounted for by the additional motion.

In terms of body size differences in the interaction, the female model consistently had a higher average TMRM (44.8°, SD=28.8) across sensors and planes than the male (37.4°, SD=34.3). On examining the start values for the forehead sensor, for example we see that the female had an average start position of -1.38° (SD=29.57), 4.85° (SD=22.71), 7.11° (SD=57.75) (sagittal, frontal, transverse) whereas the male had an average start position of -3.14° (SD=31.56), 16.2° (SD=25.62), 2.29° (SD=60.75) (sagittal, frontal, transverse). These start positions help to put the TMRM values into context as the female model had a slightly larger distance to travel to reach the end position.

On examining body size differences for the range between the maximum and minimum values, there was a three-way interaction between subject, task and body size. The female model had a larger overall average range 66.2° (SD=29.6) than the male model 59.2° (SD=33.2). These findings were interesting considering the difference in size between the models. The male model was almost double the weight and height of the female model as well as the therapists involved in the study. Verbal feedback was that it was very difficult as a team to perform the log roll on the male model. Therefore, the head Athletic Therapist may have put more focus into stabilizing the head, knowing that their team was having a difficult time controlling the body. This may have also influenced muscle activation and may not truly be reflective of a real life SCI scenario.

6.3 Electromyography

We hypothesized that the amount of muscle activation was influenced by the timing of neck realignment. Specifically, that the alignment of the neck DURING the log roll will cause less muscle activation than alignment AFTER the roll is done. The maximum muscle activation was not significant by task. It is interesting to note though that although not significant, the DURING trails had the least average peak muscle activation (10.96 %MVC, SD=17.35) and AFTER had the highest average maximum muscle activation (20.17 %MVC, SD=18.9 %MVC). When a spinal cord injury occurs, muscle activation has been known to play a role in injury exacerbation (Cusick & Yoganandan, 2002; Siegmund et al., 2009). From this study, we can infer that correcting alignment of the head during or after the roll does not cause any reflexive activation of some of the stabilizing muscles around the head and neck. The Certified Athletic Therapist can choose whichever method of timing that he or she is most comfortable with, knowing that this will not contribute to the exacerbation of the injury and secondary complications.

It was also hypothesized that the muscle activation will not be different based on the weight and size of the model. With respect to body size differences between models, peak muscle activation, there was a significant two-way interaction between subject and body size. It is possible that the head Athletic Therapist had an influence on this muscle activation for various reasons. One possible reason is that the model may not have felt completely comfortable with the hand position. For example, verbal feedback from the female model for one of the head Athletic Therapists was that in an effort to stabilize the head, the therapist was actually squeezing in on the model's ears and throat in an uncomfortable manner. However, this type of feedback in the field is not of consequence given the severity of suspected SCIs as well as additional factors, such as adrenaline and sense of urgency, that are encountered during these real life scenarios and the method in which these injuries are managed.

The male model had a lower average peak muscle activation (3.78 %MVC, SD=4.62) than the female (17.01 %MVC, SD=20.35). Although this is an interesting finding, EMG was primarily used as a tool to confirm that the subjects were acting as closely as possible to someone with a suspected severe spinal cord injury. In looking strictly at body size differences, the male did a better job staying completely relaxed than the female.

6.4 Reference Postures

EMG from the reference postures were processed and used both for calibration as well as to add context to EMG collected during the trials. These measures were mainly used to evaluate the role of model as a still model and to confirm that they did, in fact, act as if they had sustained an SCI. When we compare the values from the reference postures to the maximum values of the trials, we see that the on average maximum amplitudes do not exceed those of the reference postures. Also, the time that the peak amplitude occurred (as a percentage of the total time) was not affected by when the neck realignment took place (p=0.453).

7. Limitations

A necessary limitation to this study was the use of surface mounted equipment for data collection. Skin movement may have resulted in error and an inaccurate representation of bony movement (Adams et al., 1986). All necessary steps to reduce this error were completed such as proper skin preparation and adhesion, as well as additional taping to help minimize relative movement between the equipment and the models. Also, all of the sensors were consistently placed on the body in the exact same order and by the same person every collection. Other methods of collection such as a bone pin based marker systems or motion capture systems using optoelectronic technology (infra-red emitting or reflective markers) would not have been applicable and appropriate for this study. These types of motion capture systems would be ideal for this type of study since the markers are lightweight and very small in size (less than 1cm). However, the markers would be blocked by the Certified and student Athletic Therapists throughout most of the task, rendering the scraps of information collected useless (Boissey et al., 2011).

Despite proper skin preparation and consistent sensor placement, data were still lost. It is speculated with confidence that the addition of the lost data and/or more data collected with current methods would still yield the same results.

Another possible limitation to this study was the size of the MTx™ sensors. They were located over the spinous processes of vertebrae C4, C7, and T4, thus giving an overall motion of groupings of vertebrae rather than capturing movement directly at that spinal level. It is also possible that the instrumentation may have had an influence on the Certified Athletic Therapist's hand placement. Anecdotally, there were no reports of the influence of the instrumentation on the Certified Athletic Therapist's selected hand position and movement.

Special consideration was taken into account while collecting sEMG of the neck region because signals may have been susceptible to factors such as hard swallowing, breathing, ECG and sniffling (Sommerich et al., 2000). These factors were accounted for during the calibration process as we recorded and quantified trials of various functions that may have affected the signal to see how much (if any) muscle activity was evoked. Heartbeat contamination was removed with the use of filter which also removed a small portion of the desired signal. This was a worthwhile trade-off as the ECG dominated the bulk of the expected low levels of EMG

This study took place in a laboratory setting, which has ideal environmental conditions. During a live boarding scenario, injured victims may be outdoors in the rain and/or may have perspiration on their heads which will influence how the Certified Athletic Therapist holds the head (Boissey et al., 2011). Other factors also come in to play during a live boarding scenario such as adrenaline, fear, and noise from the crowd, which can have an influence on performance. The models being used are healthy therefore extrapolation to those with other afflictions, such as paralysis, may be difficult (Boissey et al., 2011). By monitoring and analyzing muscle activity with the use of sEMG, any muscle activation was accounted for and was negligible, and so did mimic a paralysis scenario.

8. Knowledge Generated and Future Directions

The outcome of this study affects any individuals involved in the critical initial management stages of a potential SCI and can directly be translated into current practice. Conflicting schools of thought and practice are evident in the field today, yet there has been (up to this point) minimal (if any) published evidence to suggest which timing is better. The results of this study clearly show that the timing of realignment of the head/neck complex is inconsequential. Not negating the fact that each scenario is unique, this valuable piece of information will be communicated to the Athletic Therapy community through publications in athletic therapy and sports medicine journals, provincial/national newsletters and conference presentations. Knowledge can also be transferred to Athletic Therapists through documents and/or webpages that act as quick referral guidelines for transfer techniques. Appendix A reiterates the questions and answers (Study Q & As) addressed through this thesis as well as extending beyond that and discussing out other points of knowledge that can be derived from this study.

All of the Certified Athletic Therapists used in this study were female. Another future direction should include a look at male Certified Athletic Therapists, comparing their performance to that of the females. Athletic Therapists must renew their first responder certification every third year. Therefore, these results can be incorporated into emergency care coursework. Based on the findings from this study (particularly that the subject with the least amount of experience performed the most efficiently overall) a suggestion is to ensure that first responder certification be renewed earlier than every three years or that the boarding process be integrated into regular Health Care Practitioner CPR training. By further comparing performance outcomes between groupings of Certified Athletic Therapists based on the length of time they have been certified, more insight would be gained into possible restructuring of recertification curricullae as well as timing.

Although this study used specifically Athletic Therapists, they are not the only profession that may be a first responder in a potential spinal cord injury scenario. These results from this study clearly have an impact on how suspected SCIs should be managed. Therefore, future studies should aim to compare alignment methods across and including different first responders such as paramedics, fire fighters and police officers. It is imperative that as first responders there is consistency in the realignment phase of acute SCI management.

The major focus of the study was to analyze motion from the ground onto the spine board. Other necessary future studies should include the analysis of different stages of the transfer process broken down. For example, current procedure involves sliding the athlete along the length of the board in order to center them on it. However, new techniques are emerging, such as a lateral push across the board, with no research to support this. Another example is seeing how much moving a limb that is abducted (which is a common scenario in the field) affects motion at the spine. Other future studies should include a more in depth look at hand placement during the transfer as well as equipment removal techniques.

Currently, the log roll is the only method used for prone patient transfer in Canada. Until an appropriate alternative transfer technique is developed continued research is needed to help strengthen not only the different facets of the log roll, but also to help increase its effectiveness. Finally, and most importantly, future research should aim to quantify how much movement falls within a safe range to avoid exacerbation of SCI. This crucial piece of information would help to gain insight into how the overall transfer process of an acute SCI should be managed.

9. Conclusion

The initial management stages of suspected SCIs are crucial in order to reduce the risk of secondary injury to the spine. Currently, there is a void in the literature with regards to the timing of neck realignment during the boarding process. The purpose of this thesis was to assess the movement patterns of the cervical spine and muscle activity of neck stabilizers during a prone log roll transfer specifically focussing on the timing of neck realignment. An additional purpose was to examine if the weight and the size of a victim play a role in the amount of motion and muscle activation occurring at the cervical spine during patient transfer onto a spine board. This study has demonstrated that there is no significant difference in the amount of motion generated during a prone log roll based on the timing of the neck realignment. Therefore, the therapist responsible for controlling stabilization and realignment of the head during a suspected SCI scenario can choose to correct realignment of the head at any time during the roll, provided contraindications are still adhered to. It has also shown that the weight and size of the victim may play a role in movement at the cervical spine, perhaps as a direct result of the team of therapists involved in the boarding process. The findings from this study will help enhance the knowledge of transfer techniques and will have implications on proper training techniques for those involved in the primary stages of injury management. These results can be directly transferred and used in current practice and will be communicated to the Athletic Therapy community through publications, conference presentations and possibly through an information sheet on the Canadian Athletic Therapists Association website.

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Appendix A:

Study Q & As

Take Home Message

The initial management stages of a suspected SCI are crucial. Conflicting schools of thought and practice are evident in the field today, yet there has been (up to this point) minimal (if any) published evidence to prove what the correct timing for head realignment to neutral position is better (during or after the prone log roll). This study has shown that the timing of neck realignment during the prone log roll is inconsequential in terms of the amount of motion and muscle activity. Therefore, first responders are free to correct alignment of the neck using the timing of their choice without causing further damage to a suspected SCI. Please note, additional work should be conducted, ideally with smaller sensors, to confirm the findings of this study. It was also found that the size of the victim may impact the amount of motion occurring at the neck, yet the goal still remains to stabilize the head and minimize the amount of overall motion occurring at the spine during patient transfer because currently, no method has been successful in quantifying how much motion will cause detrimental effects on SCI.

Kinematics

1. Was there a difference in the amount of motion at the cervical spine by task?
   There were no significant differences found between the levels of C4, C7, and T4 across the timing of neck realignment. However, further investigation using

smaller sensors is needed to gather more information on specific vertebral movement.

2. Was there a difference in the amount of motion at the cervical spine by weight and size of the victim? *Yes. The weight and size of the victim may play a role in the amount of motion occurring at the cervical spine during patient transfer, though more research focussing specifically on this factor is needed.*
3. Was there a difference in the amount of motion at the cervical spine by Certified Athletic Therapist? *Yes. The Certified Athletic Therapists did not perform the same despite the fact that all of them have the same academic background as well as Health Care Practitioner First Responder certification. The amount of years of certification as an Athletic Therapist may play a role in the performance of the therapist as a first responder, but more research is needed comparing not only levels of experience, but sexes of therapists.*
4. When did the least amount of motion happen? *The least amount of motion occurred during the initial stabilization phase immediately prior to realignment of the head.*
5. When did the maximum and minimum amount of motion occur timewise? *The maximum angles occurred on average midway through the trials. The minimum values occurred almost immediately following the maximum angles.*

<u>EMG</u>

1. Were the models able to perform the same across Certified Athletic Therapists? *No. One Certified Athletic Therapist in particular caused increased muscle*

activity in the female model. Verbal feedback from the model was that the hand placement of the therapist was uncomfortable, making it difficult for the model to relax. This suggests the need for future research on hand placement during stabilization so as not to invoke muscle activity which has been known to exacerbate injury.

2. Were the models able to perform the same regardless of the task? *Yes. The models had the same amount of muscle activity (which was minimal) regardless of the timing of neck realignment.*
3. Was there any reflexive EMG? *No reflexive EMG was discovered, meaning the first responder can perform the prone log roll without worry of muscle activity worsening the suspected SCI.*
4. What were the overall average activation (%MVC) and the average peak amplitude (%MVC) achieved? *The overall average activation across the trials was 2.3 %MVC and the average peak amplitude was 14.3 %MVC.*
5. When did the maximum occur timewise? *Peak muscle activation occurred at 52% of the trial length for the male and 23.4% of the trial length for the female.*

<u>Other</u>

1. Were two distinct tasks elicited? *Yes. Motion began, on average at 52% of the trial length for DURING and at 62% for the AFTER trials.*
2. What condition did CHOICE most closely resemble and were the Certified Athletic Therapists consistent across these CHOICE trials? *81% of the CHOICE*

trials were correct DURING suggesting that Certified Athletic Therapists have more of a natural tendency to correct head realignment during the prone log roll.

6. Was there a difference in the length of trials across tasks? *Yes. The CHOICE trials were the longest on average (24.25s), followed by AFTER (23.94s) and then DURING (21.94s).*

7.

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Relevant figures:

Figure 2.1.1.1: Typical characteristics of a cervical vertebra, p. 5.

Figure 2.1.2.2: Lateral view of ligaments of the cervical spine up to C3, p. 8.

Figure 2.1.2.3: Musculature of the cervical spine (lateral view), p. 10.

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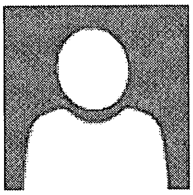

b) lateral view of the ligaments of the cervical spine

c) musculature of the cervical spine

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Figure 2.2.4: The buckling effect of the spine. *A)* The spine acts as a solid column at 30° of flexion. *B) and C)* As axial compressive loads are applied the spine begins to buckle resulting in injury *(D and E)*, p. 14.

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Figure 2.3.6: The lift-and-slide technique, p. 22.

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