Axial twist of the lumbar spine: Mechanical responses to twisted postures and potential factors for workplace injury

by

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Authors Declaration

I hereby declare that I am the sole author of this thesis. This is a true copy of the thesis, including any required final revisions, as accepted by my examiners.

I understand that my thesis may be made electronically available to the public.

ABSTRACT

While a link between magnitudes of spinal axial twist motions and the various modes of associated injury, pain reporting, and lost time claims has been tentatively established, there is need for greater investigation and understanding of the mechanical impact of axial twist motions. Researchers have compiled data sets demonstrating the relationship between twisting motions and moments and low back injury outcomes, but do not create a link to gross occupational exposures. Further, few studies can create a direct relationship between workstation design, trunk postures, and spine joint specific pain and failure mechanisms. When this limited mechanistic understanding is paired with injury prevalence statistics, they highlight a clear need to investigate the role of tissue-level axial twist exposures on occupational injury risk and workstation design guidelines to mitigate that risk.

The global objective of this research was focused on developing a relationship between working axial twist postures and intervertebral joint injury risk. The four specific questions asked were (1) What is the relationship between externally measured thoracopelvic axial twist and the actual segmental axial twist motion of the intervertebral joints? (2) Can we use ultrasound as a modality to consistently and accurately measure vertebral axial twist motion? (3) What amount of lumbar axial twist presents an elevated injury risk for working populations? (4) What movement strategies do people use to perform reaching tasks at different hand locations, and how do task parameters impact these strategies?

Study 1: Ultrasound has the potential for use to evaluate boney movement during axial twist of the lumbar spine in both *in vivo* and *in vitro* evaluations. Such segmental rotations could then be measured under controlled external thoracic axial twist conditions and in response to mechanical loading. The purpose of this study was to measure vertebral segmental rotations in a porcine model of the human lumbar spine using an ultrasound imaging protocol, and to validate use of this imaging technique with an optical motion capture system. Twelve porcine functional spinal units were fixed to a mechanical testing system, and

compression (15% of compressive tolerance), flexion-extension, and axial twist (0, 2, 4, or 6 degrees) were applied. Axial twist motion was tracked using an optical motion capture system and posterior surface ultrasound. Correlation between the two measurement systems was greater than 0.903 and absolute system error was 0.014 across all flexion-extension postures. These findings indicate that ultrasound can be used to track axial twist motion in an *in vitro* spine motion segment and has the potential for use *in vivo* to evaluate absolute intervertebral axial twist motion.

Study 2: The relationship between externally measured and internal spine axial twist motion is not well understood. Ultrasound is a validated technique (Study 1) for measurement of vertebral axial twist motion and has the potential for measuring segmental vertebral axial twist *in vivo*. The purpose of this study was to evaluate lumbar segmental axial twist in relation to external thoracopelvic twist using an ultrasound imaging technique. Sixteen participants kneeled in a custom-built axial twist jig which isolated motion to the lumbar spine. Participants twisted from neutral to 75% of maximum twist range of motion in an upright flexion-extension posture. Thoracopelvic motion was recorded with a motion capture system and L1 to S1 vertebral axial twist was recorded using ultrasound. Maximum thoracopelvic axial twist motion was 41.1 degrees. The majority of axial twist motion occurred at the L2-L3 (46.8% of lumbar axial twist motion) and L5-S1 (33.5%) intervertebral joints. Linear regression fits linking axial twist at each vertebral level to thoracopelvic axial twist ranged from 0.43 to 0.79. These findings demonstrate a mathematical relationship between internal and external axial twist motion, and suggest that classic use of L4-L5 to represent lumbar spine motion may not be appropriate for axial twist modeling approaches.

Study 3: Axial twisting exposures have been repeatedly identified as a risk factor for occupational low back pain and injury, but there is a need for an improved understanding of the role of axial twist magnitude and associated moment as modifiers of the cumulative load tolerance of intervertebral joints.

The purpose of this study was to mathematically characterize the relationship between axial twist motion magnitudes and the cumulative load tolerance of porcine cervical functional spinal units. Twenty-four porcine functional spinal units were fixed in a mechanical testing system under compressive load (15% of compressive tolerance) and in a neutral flexion-extension posture. Specimens were axially twisted to 5, 7.5, 10, 12.5, 15 or 17.5 degrees at 1 Hz until failure or 21 600 total cycles. Cumulative applied axial twist was recorded, and exponential functions were fit to the twist magnitude-cumulative twist moment recordings. Weighting-factor functions for cumulative axial twist moment injury risk were developed based on absolute axial twist magnitude and twist normalized to maximum range of motion. The non-linear weighting-factors have potential use in assessment of cumulative axial twist injury risk in occupational tasks.

Study 4: The magnitude of axial twist in the lumbar spine in relation to reaching tasks is currently unknown. Therefore, the purpose of this study was to investigate lumbar spine axial twist during simulated occupational tasks across a range of forward and lateral reach distances, task heights, and exertion directions. Twenty-four participants performed single-handed, right-handed exertions against a load cell in three directions (upward, downward, forward push), at two heights (shoulder, elbow), and at 11 different hand target locations corresponding to current ergonomic reach guidelines. Thoracopelvic and right upper limb postures were recorded using an optical motion capture system, and trunk muscle activation was recorded using surface electromyography. Participants performed a contralateral twist at both the thoracopelvic spine and pelvis about the feet for directly forward hand targets, and twisted up to 19.9 degrees and 12.1 degrees at the lumbar spine and pelvis, respectively, at the most lateral hand target locations. Lumbar flexion and shoulder elevation each increased with reach distance to a maximum of 5.6 degrees and 64.9 degrees, respectively, at the furthest, directly forward hand target location. Hip and abdominal muscle activation exceeded 10% MVC for the most lateral hand target locations, and exhibited the highest activation for upward and forward push exertions. These findings suggest that future ergonomics guidelines should assess reaching and exertion tasks to hand target locations beyond 60-degrees from the midline of the body and consider them as non-optimal zones.

The collection of studies in this thesis was structured to improve current ergonomics reach guidelines and provide a physiological and biomechanical basis for reach distance recommendations incorporating the low back. The findings from these studies have important implications for researchers, ergonomists, and clinicians assessing injury risk related to twisted occupational postures.

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Last, I would like to send out a heart-felt thank you to all of my friends and colleagues that chose this same path before me or after me. The appreciation of what it takes to get here is something we will always share. I look forward to staying friends far beyond the end of our careers. Keep in touch.

Authorship Contribution Acknowledgements

I hereby declare that my contribution to each journal publication that was produced from my thesis (currently in press, accepted, and submitted) was from my own work, and for each study that I was responsible for the conception and/or design, preparation, collection and analysis of the data, and the writing and editing of the manuscripts.

I hereby declare that the contribution of author Dr. Jack P. Callaghan on each manuscript was to the conception and/or design, editing of the manuscripts, assisting with instrumentation, and securing external funding for the research.

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CHAPTER 1
INTRODUCTION

1.0 Overview

While a link between magnitudes of spinal axial twist motions and the various modes of associated injury, pain reporting, and lost time claims has been tentatively established, there is need for greater investigation and understanding of the mechanical impact of twist motions. Within Canada, over 10% of the adult population has been disabled by low back pain in the last 6 months (Cassidy et al., 1998) and over 18% of all Ontario lost time claims are due to low back injury (WSIB, 2014). Time spent in twisted postures and the frequency of twisted postures have both been shown to induce low back injury resulting in lost time claims (Hoogendoorn et al., 2002; van den Heuvel et al., 2004). While researchers have compiled data sets demonstrating the relationship between twisting motions and moments and low back injury risk, these are most often under specific, controlled conditions, and do not create a link to gross occupational exposures. Further, few studies can create a direct link between workstation design, gross trunk postures, and spine joint specific pain and failure mechanisms. When this limited mechanistic understanding is paired with injury prevalence statistics, they highlight a clear need to investigate the role of tissue-level axial twist exposures on occupational injury risk and workstation design guidelines to mitigate that risk.

Previous work has identified inconsistencies with naming and terminology in relation to motion and loading of the lumbar spine (Drake, 2008). The terms torsion, twisting, and rotation are often ambiguously used when referring to the motion of a functional lumbar spinal unit, the applied load or causal moment that created motion, and the passive resistive load induced by that motion. Each of these parameters are intimately linked, but distinction of their dependence and causal role in a given scenario is important to clearly identify. In this document, spine motion will be referred to as axial twist motion, and spine moment will be referred to as axial twist moment.

1.1 The Low Back Pain Problem

Low back pain (LBP) and associated disorders are the most common and influential musculoskeletal issues in and out of the workplace. Four of every five adults have some form of back pain during their life (Cassidy et al., 1998; Kelsey & White, 1980), and an estimated 14.5 million people in Canada experience an acute low back pain episode annually (Beaudet et al., 2013). A recent systematic review illustrated these numbers on a global scale to correspond to approximately a 12% point prevalence of low back pain lasting greater than one day and a 23% one-month prevalence (Figure 1.1). Prevalence was highest during middle age, which has been referred to as the most productive years in a working life (Hoy et al., 2012). This equates to an annual back pain prevalence ranging from 15% to 45%, and back pain has become the most common cause of activity limitation in adults less than 45 years old (Marras, 2000).

More specifically, occupational axial twisting motions have been identified as a significant contributor to occupational low back disorders (Chaffin et al., 1991; Kelsey et al., 1984; Marras et al., 1993; Punnett et al., 1991). Axial twisting motions have been reported to be associated with one-third of occupational low back disorders (Marras et al., 1998) and nearly 20% of worker compensation costs (Snook, 1978). While this problem is quite apparent, the cause is much more complicated to explain and identify. Low back pain describes the result of damage-, degeneration-, or impingement-induced pain pathways that can be initiated by a wide range of risk factors. Mechanical loads, psychophysical modulators, and occupational factors each play a role in the development of low back pain, and extensive research is required on each of these areas in an attempt to explain and prevent onset of occupational low back pain.

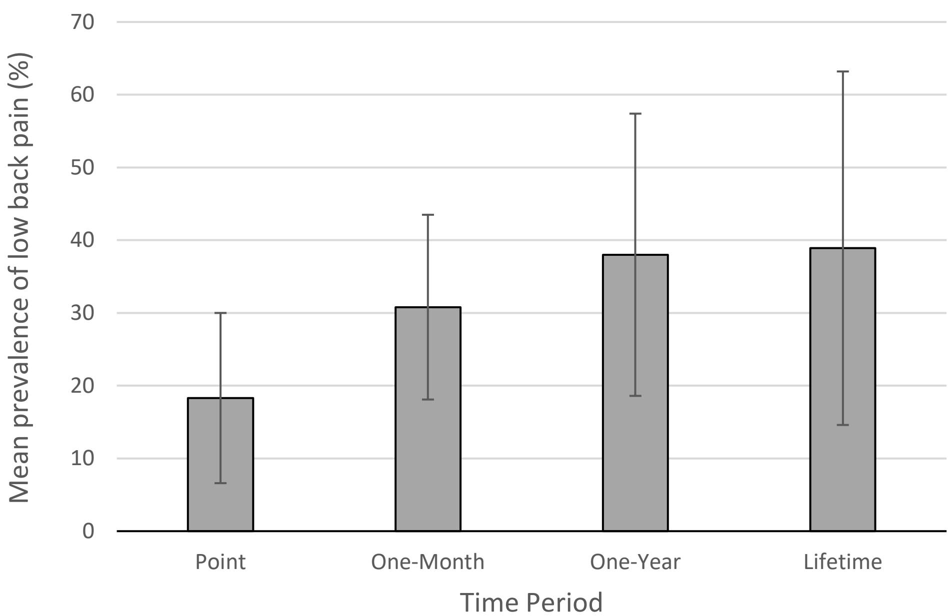


Figure 1.1: Mean prevalence of low back pain according to prevalence period. Significant differences were present between point, one-month, and one-year prevalence. No difference was present between one-year and lifetime prevalence. Error bars represent ± 1 standard deviation. Adapted from Hoy et al. (2012).

Epidemiological Evidence of Axial Twist Injury

Axial twisting of the spine is commonly identified as a contributor to low back pain or disorder, and several studies have identified twist exposures as a significant risk factor for industrial and occupational low back disorders (Bernard, 1997; Hoogendoorn et al., 1999, 2002; Marras, 2000; van den Heuvel et al., 2004; Videman et al., 2005). Twist motions, whether required or not, are a very common strategy to reach hand target locations during work ranging from precision tasks to high load lifting. In general, these types of movements are performed on a daily basis and often repetitively in many industrial environments.

Historically, manual materials handling activities, especially lifting, had been thought to dominate low back disorder risk, and received a large proportion of the attention in epidemiological and biomechanical research (Kelsey et al., 1984; Reid & Costigan, 1987; Sommerich et al., 1993). Concurrent and subsequent research investigated factors coupled with lifting exposures, and identified static postures (Kelsey, 1975; Magora, 1970, 1973), movement frequency (Andersson, 1999; Kelsey et al., 1984;

Keyserling et al., 1988), exertion direction (Chaffin & Park, 1973; Kelsey & White, 1980; Kelsey, 1975; Kelsey et al., 1984), and repetition (Kelsey et al., 1984; Rubin et al., 1987) as risk factors for developing low back disorders. In an extensive study of over 400 jobs across 48 industries, Marras *et al*. (1993) found lifting frequency, load moment, trunk lateral velocity, trunk twisting velocity, and trunk sagittal angle as the most influential parameters to distinguish between high and low risk of low back disorders. In automotive assembly workers, trunk flexion and twisting motions have been directly associated with low back disorders, while the combination and increased duration of these movements increased injury reporting risk (Punnett et al., 1991). Further, axial twist of the torso coupled with moderate load induced a three-fold increase in intervertebral disc herniation occurrence (Kelsey et al., 1984).

Pain Generation

While the risk factors associated with axial twist motions and combined loads are well-defined, modern research continues to investigate the link between axial twist exposures and specific tissue damage. As will be discussed in Chapter 2, twist exposures can lead to damage of the facet joints (McCall et al., 1976; Schwarzer et al., 1994; Schwarzer, Wang, Bogduk, et al., 1995) and the intervertebral discs (Boos et al., 2000; Ito et al., 1998). The link between tissue damage and pain onset is less clear as damage or degeneration of the intervertebral disc does not always coincide with pain onset (Boden et al., 1996; Jensen et al., 1994), and painful facets cannot always be linked to detectable joint damage (Schwarzer, Wang, O'Driscoll, et al., 1995). Pain-provocation studies have highlighted the facet joints as frequent sites of low back pain (Adams, 2004; Schwarzer, Wang, Bogduk, et al., 1995) and revealed the role of the facet joints in pain generation by both causing pain and removing pain. A large study of conscious patients undergoing surgery for herniated disc or spinal stenosis showed sharp, localized facet joint pain in approximately one-third of patients when the facet joint was mechanically provoked (Kuslich et al., 1991).

Schwarzer *et al*. (1995) reported that when facet joints were injected with local anesthetic, 15% of patients saw a significant reduction in facet joint pain.

While these and several other studies have identified the facet joints as a frequent cause of low back pain, detailed patient histories have not yet yielded any consistent clinical predictors of such pain (Adams, 2004; Schwarzer, Wang, Bogduk, et al., 1995). Axial twist moments and associated joint kinematics need to be explored to gain insight into the structural changes that occur both as a result of injury or degeneration and as a precursor to such conditions.

1.2 Personal Factors That May Influence Low Back Pain

Pain generation and the underlying mechanism tends to be studied from a mechanical point of view, but it is important to investigate personal factors that may predict onset of low back pain. Some factors are out of the control of a worker or participant, such as age or gender, while others are controllable or are a response to other occupational factors such as working environment. Predictive studies have shown physical, personal, and psychological factors all to be somewhat correlated with onset of low back pain, but some of these relationships are confounded or weak at best.

Personal physical factors such as low physical activity levels (Thomas et al., 1999), smoking (Power et al., 2001; Thomas et al., 1999), and lower leg pain (Cherkin et al., 1994) are commonly investigated due to the ability to connect them with an underlying injury or pain mechanism. Smoking may be an indirect factor related to overall physical well-being, and is most likely due to increased symptom reporting rather than actual increases in related lost time claims (Ferguson & Marras, 1997; Marras, 2000). Numerous studies have investigated previous history of low back pain (Bigos et al., 1986; Chaffin, 1974; Punnett et al., 1991; Vällfors, 1985) and most studies (87%) identify it as the best predictor of risk (Ferguson & Marras, 1997; Marras, 2000). Physical factors are often linked to tissue health and loading history and act as modifiers to tissue tolerance or load capacity.

Age and gender play a key role in predicting low back pain (Andersson, 1999; Cherkin et al., 1994; Pincus et al., 2002; Thomas et al., 1999) with the highest frequency of symptoms between the ages of 35 and 55, and peak risk occurring at 40 years for men and 50-60 years for women.

Psychological predictors of low back-related lost time such as depression (Cherkin et al., 1994; Pincus et al., 2002) and dissatisfaction with employment (Thomas et al., 1999) remain somewhat controversial, as an associated low back pain mechanism cannot be fully explained and the validity and reliability of the underlying data are uncertain (Andersson, 1999). Low back pain often becomes a convenient diagnosis for workers who are disabled for socioeconomic, work-related, or psychological reasons (Andersson, 1999). Regardless, it is import to understand the role of all personal factors in low back pain development and their interaction with more explainable occupational and physical factors.

1.3 Occupational Factors That Exacerbate Low Back Pain

Numerous epidemiological studies have been published to identify common workplace musculoskeletal disorder (WMSD) risk factors in various workplaces and work types. The goal of such studies is to isolate specific biomechanical aspects of work that contribute to worker pain and injury in the modern work era. Critical reviews seek to summarize and evaluate the quality of such studies, and have identified five factors common to most epidemiological studies: 1) heavy physical work, 2) lifting and forceful movements, 3) bending and twisting, 4) whole body vibration, and 5) static postures (Bernard, 1997; Hoogendoorn et al., 1999; Marras, 2000). These reviews found strong evidence that linked risk of low back pain to lifting and forceful movements, whole body vibration, and bending and twisting. The strong link between low back pain with bending and twisting relates directly to the injury prevalence that contributed to the development of this thesis. The risk related to bending and twisting needs to be separated to be able to properly identify and mitigate specific actions or tasks that increase WMSD risk associated with these often coupled but orthogonal motion axes. Moderate evidence was available supporting risk related to heavy physical work, and tended to have more information on the underlying load-based injury mechanism. While static work postures are commonly cited as an occupational risk factor, they lack evidence specifically related to injury risk (Bernard, 1997; Marras, 2000).

Epidemiologic studies typically evaluate outcomes related to specific work factors, but they have low sensitivity to show a quantitative relationship between these parameters. Further, the outcomes of these studies can vary based on the dependent measures the researchers observed (Figure 1.2), which makes it difficult to link outcomes to particular risk factors and to distinguish between symptoms and injuries (Marras, 2000). As a result, a few large-scale biomechanical field studies have evaluated numerous jobs and industries, and defined direct relationships between biomechanical risk factors and risk of low back disorder outcomes. With a focus on spine motion and loading, Marras *et al*., (1993) and Marras *et al*. (1995) used a lumbar motion monitor to quantify physical exposures in more than 400 jobs. From this evaluation, five factors could be used to describe the relationship between incidence of low back disorder and lost or restricted time due to low back disorder: lifting frequency, sagittal torso flexion angle, lateral velocity, twisting velocity, and external load moment (Marras, 2000). Norman *et al*. (1998) performed a case-control risk evaluation on over 200 workers selected from a pool of over 10 000 available employees. This study evaluated the importance of modelled 2D peak spine loads, hand loads, trunk kinematics and cumulative spine loads as predictors of low back disorders and found five factors with significant odds ratios: integrated lumbar moment, hand force, peak L4-5 shear force, and peak trunk velocity. Compression was also highly correlated with these parameters and could be substituted for moment or shear without altering the odds ratios. The combination effect of these risk factors amplified risk, as workers in the top 25% of exposure to all risk factors had a six-fold increase in risk compared to those in the bottom 25% (Norman et al., 1998). While these identified factors are still relevant, the planar nature of this modelling approach did not allow for low back axial twist analysis.

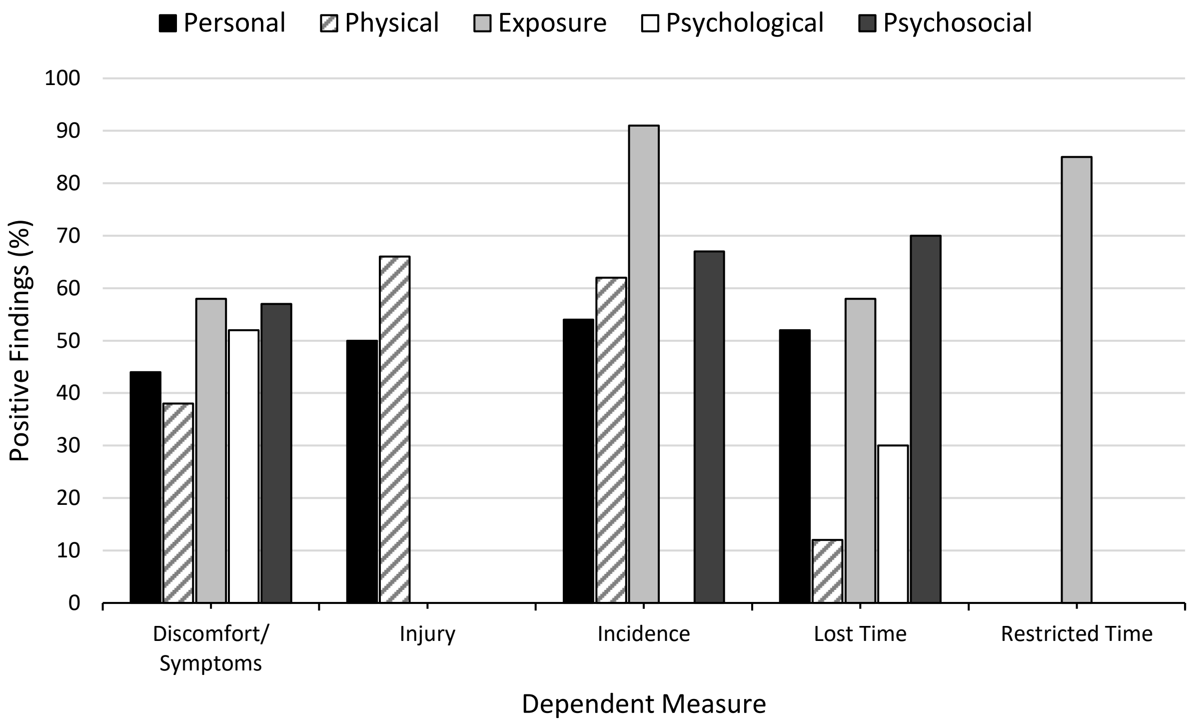


Figure 1.2: Positive findings for a relationship between low back disorder risk across a range of epidemiologic studies and surveillance measures (by risk factor type). Note each risk factor type is greater when determined from incidence than from symptoms or discomfort. This indicates potential false negatives when evaluating symptoms only. Adapted from Marras (2000).

While epidemiologic studies can identify correlations between exposures and outcomes, biomechanical studies provide the specific exposure-level detail required to drive workplace interventions and risk prevention strategies. Other research aims to target specific risk factors and evaluate how exposure changes can affect both injury risk and specific biomechanical responses. For example, gross movement and work research targeting lifting exposures found that jobs requiring lifting more than 110 N more than 25 times per day had 3 times the risk of acute prolapsed lumbar IVD than people not lifting this weight at this repetition rate (Kelsey et al., 1984). Further, when these lifts included coupled twisting exposures, this elevated risk was present at frequencies as low as 5 times per day (Kelsey et al., 1984). Other joint or functional motion segment specific research has found kinematic and kinetic changes in the lumbar spine in response to axial twist posture changes (Drake, 2008; Drake et al., 2005). This wide range of research is necessary to identify risk and injury outcomes at a body and a segment or joint level, and also to explain the mechanism for injury at a tissue or joint level.

1.4 Lumbar Spine Axial Twist Motions and Facet Function

Spinal axial twist motions are extremely common in modern industrial work environments, and have been highlighted as a key risk factor for low back disorders. The moment, velocity, and rotational stiffness associated with such tasks has been investigated during standing (Marras et al., 1993; McGill & Hoodless, 1990) and sitting exposures (Kumar et al., 1996). Both *in vivo* and *in vitro* research have clearly identified the importance of the facet joints during twisting motions and the effect lumbopelvic posture has on axial twist range of motion (Adams & Dolan, 1995; Rohlmann et al., 2001). One of the functions of the facet joints is to protect the disc from excessive axial rotation in the lumbar spine (Adams & Hutton, 1981; Ahmed et al., 1990; Duncan & Ahmed, 1991) and this resistance to load is augmented by the compression due to body weight (Farfan et al., 1970). The lumbar spine twist range of motion is a function of the amount of spine flexion (Drake & Callaghan, 2008; Haberl et al., 2004) and the associated gap between the articular surfaces of the facet joints (Haberl et al., 2004). When flexed, the lumbar motion segments exhibit greater twist range of motion (Figure 1.3) with modified rotational stiffness (Drake & Callaghan, 2008), and the load-bearing capacity of the facet joints is decreased (Drake et al., 2005; Marras & Granata, 1995). This postural mechanism has been hypothesized to be a result of facet joint disengagement and increased distance between articular surfaces prior to rotation (Drake & Callaghan, 2008). Little or no research has been able to medically image and evaluate facet joint geometry changes during physiologic movements in neutral, flexed and extended postures. Further, the direct link between facet joint geometry changes, vertebral segmental rotations, and gross torso rotations observed in the workplace remains unknown.

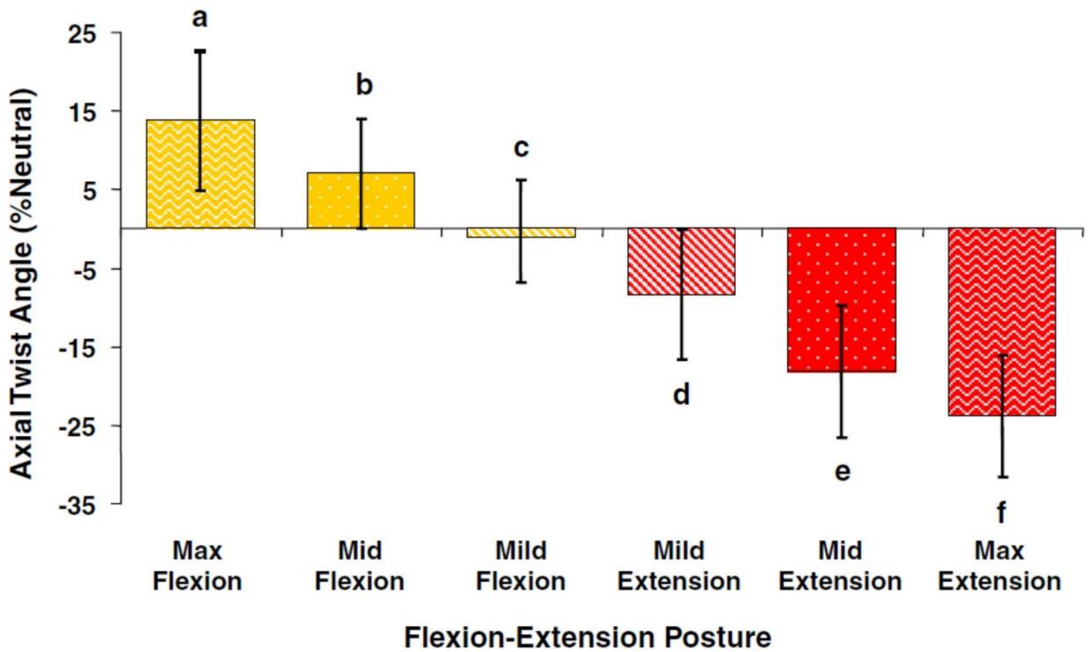


Figure 1.3: Normalized axial twist angles while in a range of flexion and extension postures. Twist angle was greater than neutral while flexed and less than neutral while extended. Adapted from Drake & Callaghan (2008).

1.5 Loading Parameters

Much of the literature on occupational exposures highlights twisting as a key risk factor for low back disorders, but the specific influence of frequency and magnitude characteristics needed to make specific workplace interventions remain unclear and scarce. The posture of an intervertebral joint has a great effect on joint kinematics (Adams & Hutton, 1983; Panjabi et al., 1983), load distribution (Ahmed et al., 1990), and failure mechanism (Drake et al., 2005; Veres et al., 2010). Repetitive loading has been established as a key factor and has consistently achieved intervertebral disc failure (Callaghan & McGill, 2001a; Drake et al., 2005; Howarth & Callaghan, 2012; Marshall & McGill, 2010). Cadaveric and *in vitro* models of the human lumbar spine have successfully and extensively described specific pathways for injury (specifically herniation) progression, annulus fibrosus delamination characteristics, disc volume and fluid content, and several other tissue properties under these conditions. These multi-axis *in vitro* loading parameters all should be considered and should attempt to mimic *in vivo* loading in order to properly model an axial twist-induced injury mechanism.

1.5.1 Axial Twist Motion and Axial Twist Moment

The terms axial twist and axial moment (or synonymous terms) are most often combined when describing injury risk factors, especially when related to occupational controls. The terms are intrinsically dependent since an active (i.e. muscular) moment creates an axial twist motion and a passively applied axial twist motion creates a resistant moment. Some research has found maximum moments before failure between 8.5-32.7 N·m with various magnitudes of secondary axis (compression) loading and disc degeneration (Adams & Hutton, 1981; Farfan, 1969; Krismer et al., 1996; Pearcy & Hindle, 1991). Other studies investigating combined rotational and sagittal loading have attempted to partition the applied loads (maximal or submaximal) between the intervertebral disc and the facet joints (Adams & Hutton, 1983). These act as a guide for modelling joint load capacity and total load on the intervertebral joint, but seldom give detailed reports on the associated twist magnitudes. Drake & Callaghan (2008) observed greater lumbar passive axial twist and less lumbar stiffness when coupled with maximum flexion when compared to an upright spine posture. While these extreme postures can be linked to increased risk of occupational low back disorders, the details to define a threshold between “safe” and “injury-inducing” magnitudes remain unclear. Facet joint changes are a function of twist magnitude (Haberl et al., 2004), but have not been directly linked to applied moment. Investigations of combined loading using fixed moment applications provide important information to model the human lumbar spine, but do not necessarily help define the threshold between safe and injury-inducing exposures. As discussed, axial twist moments and axial twist motions are inherently dependent, but to relate injury mechanisms to physiological twist exposures, tissue-level modelling also needs to be evaluated from the perspective of a range of applied axial twist motions. Further, modern ergonomics practices intend to design workstations to fit workers.

Despite this, many industrial workstations have fixed task components and force workers to achieve a given posture, not a given load. Both workstation guidelines and ergonomics interventions are driven by externally measured postures. The relationship between these postures and injury progression needs to be developed from the perspective of trunk motions and associated internal axial twist moments to create and update guidelines for twisted work exposure limits and reach dimensions that match tissue-level injury mechanisms.

1.6 Thesis Structure and Research Questions

This thesis is presented as three introductory chapters followed by four chapters representing four specific investigations related to axial twist of the lumbar spine and potential factors for workplace injuries. This chapter (Chapter 1) presents an overview of axial twist as a problem related to occupational exposures, personal factors, and basic function and loading parameters of the twisted lumbar spine. The following two chapters (Chapter 2 and 3) present common *in vitro* and *in vivo* literature reviews outlining relevant material to understand the mechanics and injury potential of the lumbar spine during axial twist motions and loading. The study chapters (Chapters 4-7) outline two *in vitro* and two *in vivo* investigations. Each chapter is presented as an introduction with an expanded relevant specific literature review, methodology, detailed results, discussion and conclusion. The final content chapter, Chapter 8, provides a global discussion of the thesis goals, contributions and outcomes. A bibliography of all references is presented at the end of this document (Chapter 9). The following sections present the global thesis research questions followed by a brief summary of each investigation and the associated research questions and hypotheses.

1.6.1 Global Thesis Questions

1. What is the relationship between externally measured thoracopelvic axial twist and the actual segmental axial twist motion of the intervertebral joints?

2. Can we use ultrasound as a modality to consistently and accurately measure vertebral axial twist motion?

3. What amount of lumbar axial twist presents an elevated injury risk for working populations?

4. What movement strategies do people use to perform reaching tasks at different hand locations, and how do task parameters impact these strategies?

1.6.2 Global Thesis Flow Chart

This thesis is designed to answer the preceding global questions and build the relationship between working postures used in industrial tasks and lumbar axial twist injury risk by way of developing a relationship between trunk postures and intervertebral motion. These links work together to develop suggested improvements to current ergonomics guidelines that could reduce injury-causing lumbar axial twist postures. Study 4 (Chapter 7) directly measured working postures and loading strategies during simulated work, Study 3 (Chapter 6) estimated injury risk in an intervertebral joint model based on axial twist magnitude, Study 1 (Chapter 4) developed and validated an ultrasound technique to measure intervertebral twist motion, and Study 2 (Chapter 5) used that technique to directly link gross thoracopelvic motion to individual intervertebral joint motion. The linkages between each of four independent investigations are illustrated in Figure 1.4. Each thesis study is described in general (1.6.3) and in detail (Chapters 4-7) in this thesis.

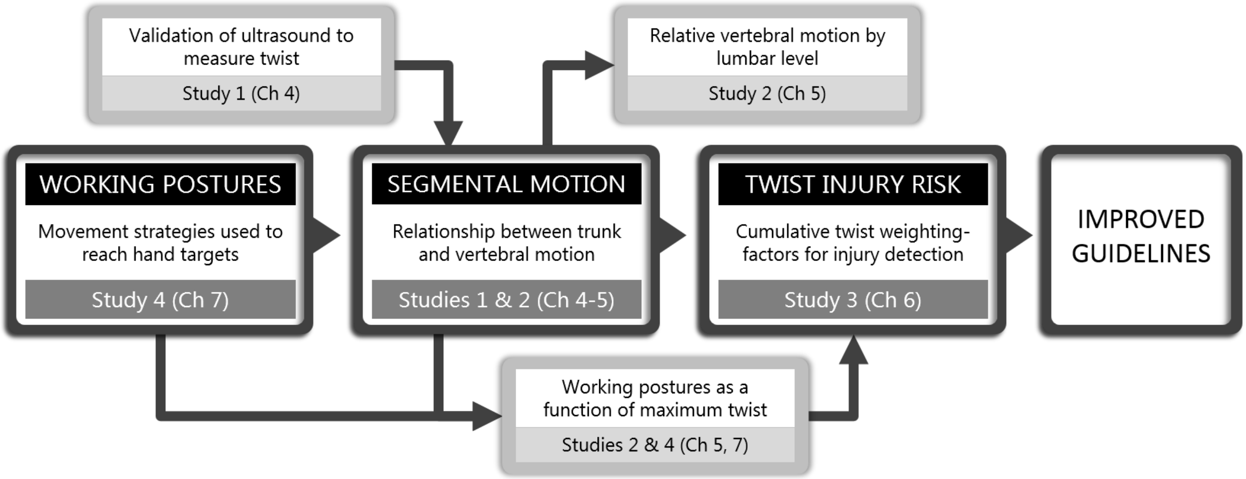


Figure 1.4: Studies 1-4 link together to develop a strategy for improved ergonomics reach envelope guidelines.

1.6.3 Specific Thesis Studies

All hypotheses for specific studies are stated as *alternative hypotheses* (H_1). Explanation and justification are provided following the statement of each hypothesis.

Study #1 (in vitro): This study provided validation of an ultrasound method to measure segmental vertebral axial twist in comparison to vertebral axial twist measured by a low-error optical motion tracking system.

1. Are ultrasound measures of vertebral axial twist valid compared to motion capture measures?
2. Are between-system differences affected by specimen flexion-extension posture?

Study 1 Hypotheses

1. *H_1: Agreement between the two intervertebral angle measurement systems is similar across the evaluated axial twist range of motion.*
2. *H_1: Ultrasound and motion capture intervertebral angle measures show equal agreement in neutral, flexed, and extended sagittal plane postures.*
    - Identical points were tracked for both systems regardless of posture. Both systems were expected to accurately track intervertebral motion across the evaluated range of motion and flexion-extension postures in this study.

Study #2 (in vivo): This study used an ultrasound imaging technique validated in Study 1 to relate *in vivo* segmental vertebral axial twist to external thoracopelvic axial twist at each lumbar vertebral level. This study presented the contribution of each vertebral level to whole lumbar spine motion during maximum passive twist range of motion.

1. What is the relationship between lumbar (aggregate) vertebral twist and externally measured thoracopelvic twist?

2. How is lumbar spine twist partitioned across each vertebral level?

Study 2 Hypotheses

1. *H_1: Intervertebral angles increase moving from inferior to superior joints in the lumbar spine.*

    - MRI and radiography studies *in vivo* have similarly shown increased intervertebral angle from L1-L2 to L4-L5 when an external rotation is applied (Haughton et al., 2002; Pearcy & Tibrewal, 1984a).

2. *H_1: Thoracopelvic axial twist is greater than total lumbar spine axial twist.*

    - Externally measured thoracopelvic angle was expected to be greater than the total lumbar axial twist angle, measured as L1 angle about the pelvis. Several studies have shown similar findings with aggregate lumbar angle accounting for 20% of externally measured rotation range of motion in MRI and CT images of human participants (Haughton et al., 2002; Ochia et al., 2006). Further, the thoracic spine has been suggested to produce more movement in twist than the lumbar spine (Fujii et al., 2007), and can account for over 40 degrees of axial twist during maximal thoracopelvic rotation tasks (Willems et al., 1996).

Study #3 (in vitro): This study characterized the relationship between axial twist motion and moment magnitudes and the cumulative load tolerance of porcine cervical spinal units. From this relationship, non-linear equations for deriving appropriate weighting-factors for estimates of cumulative axial twist exposure were developed.

1. What is the effect of externally applied twist magnitude on number of cycles to ultimate failure?

2. Does mode of failure change with increased magnitude of axial twist (disc-driven to facet-driven injury mechanism)?

Study 3 Hypothesis

1. *H_1: Cumulative axial twist moment tolerance decreases with increasing axial twist magnitude.*

   - Cumulative axial twist moment tolerance was expected to increase with lower axial twist magnitudes. Non-linear relationships between cumulative load tolerance at failure and loading magnitude have developed power function relationships for both compression and shear loading (Howarth & Callaghan, 2013; Parkinson & Callaghan, 2007b). A similar relationship was expected with decreased cumulative load tolerance at failure with increased magnitudes of applied axial twist.

Study #4 (in vivo): This study characterized posture and movement strategies during simulated industrial relevant tasks that induce forward and lateral reach and associated lumbar axial twist. Trunk and right upper limb kinematics and selected trunk muscular activation data were used to assess low back injury risk potential related to reach distance and hand target location.

1. What effects do forward and lateral reach distances have on low back muscular demands during axial twist exposures?

2. What lumbar axial twist movement strategies do participants use to perform forward and lateral reaching work?

Study 4 Hypotheses

1. *H_1: Lumbar axial twist increases with increased laterality of hand target location.*

    - Lumbar spine (thoracopelvic) axial twist angle was expected to increase at more lateral hand target locations, regardless of reach distance. Perceived shoulder discomfort increases with deviation from mid-range reaches at approximately chest height (Dickerson et al., 2006), and participants use postural strategies to achieve working postures that minimize upper limb discomfort (Babski-Reeves et al., 2005; Helander & Zhang, 1997). It was expected that lumbar axial twist would be the primary strategy to reach lateral hand target locations, rather than upper limb strategies.

2. *H_1: All recorded muscles show increased activation with increased lumbar axial twist.*

    - With increased axial twist, muscle activation and co-activation are expected to increase. Trunk muscle co-activation has been shown to increase during twisting motions (McGill, 1991), and trunk muscle activation level increases when coupled with flexion and lateral bending (Marras et al., 1998). It was expected that the activation of the recorded muscles would increase with increasing laterality of hand target location.

CHAPTER 2

GENERAL LITERATURE REVIEW FOR *IN VITRO* STUDIES

2.1 Overview

Two of the studies in this thesis (Study 1 – Chapter 4; Study 3 – Chapter 6) use an *in vitro* porcine cervical spine model as a surrogate for the human lumbar spine. Much of the experimental design and protocol details are similar between the two studies and are guided by decades of *in vitro* research on human spine biomechanics. This chapter outlines the anatomy of the human lumbar spine, sources of pain, and the movement and mechanics of both healthy and injured spines. Details of similar animal and human spine models are discussed in terms of specimen storage, loading, and response modulators relevant to the studies in this thesis. Specific details not common to both *in vitro* studies are included in the background and introductory information provided with the appropriate study chapter.

2.2 Lumbar Spine Structure and Pain Mechanics

2.2.1 Functional Spinal Segment Anatomy

The lumbar spine exhibits motion and deformation in very close proximity to neural, vascular and musculoskeletal structures during movement and loading. Neural roots branch off the spinal cord through the intervertebral foramina and then further branch to innervate the annulus fibrosus and intervertebral facet joints (Figure 2.1). Figure 2.2 and Figure 2.3 illustrate the closeness of neural and vascular structures to the boney anatomy and the disc and ligament elements of the lumbar spine. Both normal and abnormal lumbar spine motion have great potential for mechanical compression of these tissues which can lead to disruption of several structures along the neural pathway. Investigation of the changes in distance between these structures during movement is essential to understanding mechanisms for pain and injury in the lumbar spine.

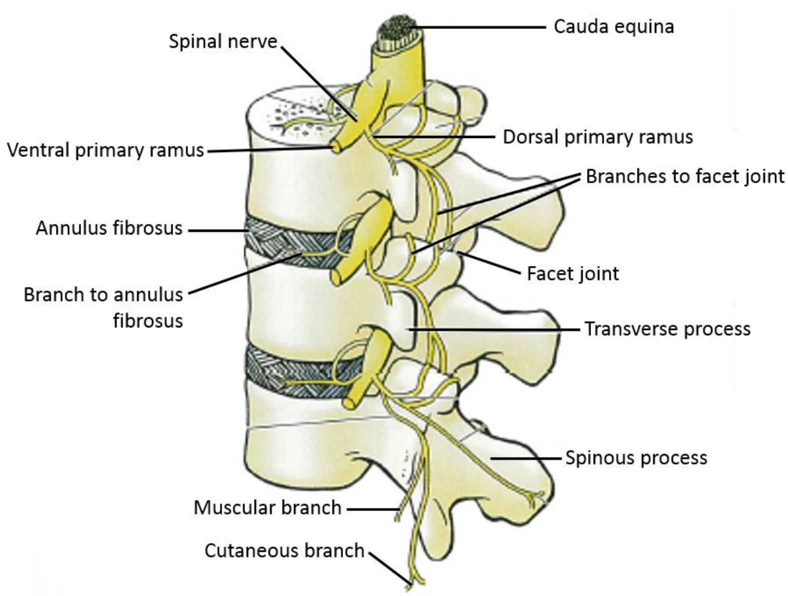


Figure 2.1: Spinal nerves branching off the spinal cord through the intervertebral foramina and further branching to innervate disc and joint components. Figure adapted from Moore, Dalley, & Agur (2013), p.506.

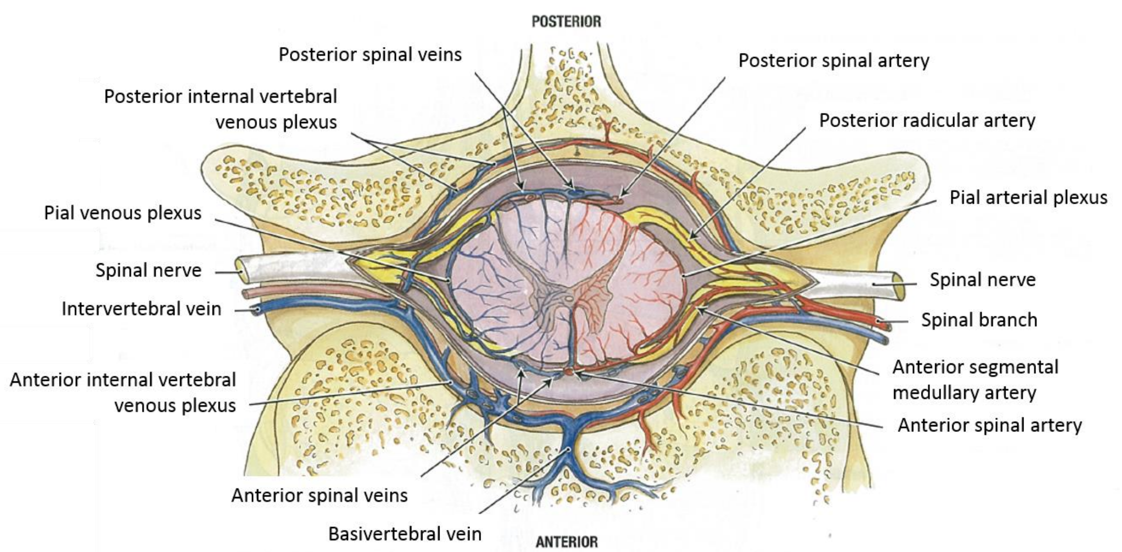


Figure 2.2: A transverse section of a vertebrae showing the proximity of neural and vascular structures to the boney surfaces. The nerves, arteries and veins are illustrated in yellow, red, and blue, respectively. Figure adapted from Agur & Dalley (2013), p. 343.

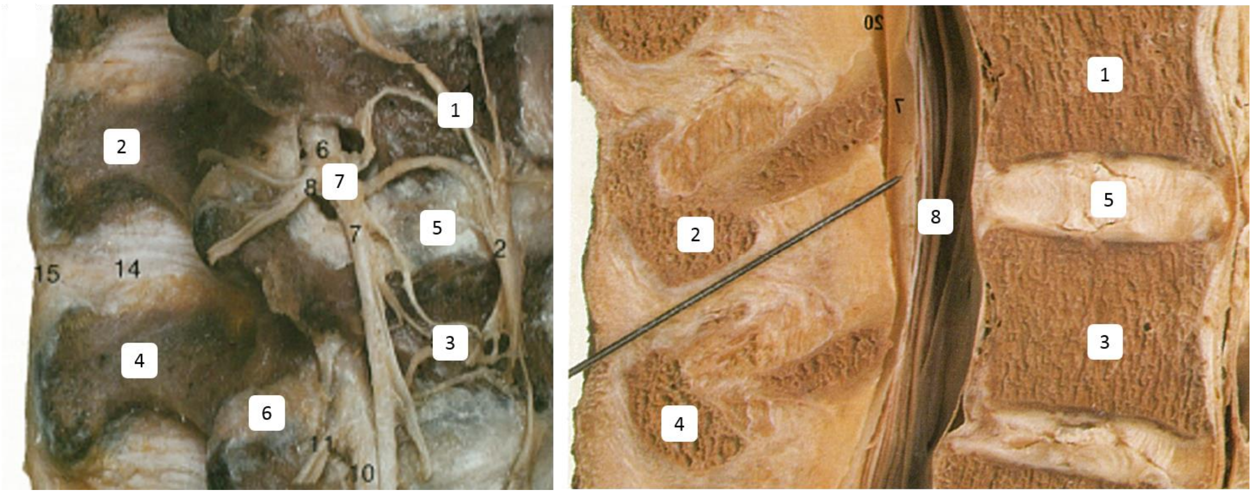


Figure 2.3: A) Dissection of the upper lumbar region, and B) an analogous right side sagittal section. Numbered labels identify the 1) superior vertebral body, 2) superior spinous process, 3) inferior vertebral body, 4) inferior spinous process, 5) intervertebral disc, 6) facet joint, 7) nerve root emerging from the intervertebral foramen, and 8) descending nerve roots of the cauda equina. Figure adapted from McMinn & Hutchings (1988), p. 85 and McMinn, Hutchings, & Logan (1984), p. 49.

2.2.2 Nerve Root Compression

The prominent role of posterior spinal articulations in pain generation has been apparent for over half a century with early research linking sciatic pain to lumbar nerve root compression and narrowing of vertebral canals (Epstein et al., 1962; Smyth & Wright, 1958; Verbiest, 1954). The presence of dense adhesions between the foramen and nerve root (Goddard & Reid, 1965) and multiple sources of foramen narrowing (Epstein & Epstein, 1959; Epstein et al., 1962; Friedmann, 1961; Schnitker & Curtzwiler, 1957) indicate the possibility for compression or mechanical irritation of the nerve roots in the lumbar spine (Panjabi et al., 1983). While this early research indicated a strong link between compression and pain, modern studies have confirmed neural tissue compression as a pain generating pathway (Hubbard et al., 2008; Winkelstein & DeLeo, 2004).

Work following the identification of mechanical irritation attempted to isolate factors that may increase the risk of this compression and relate its onset in both degeneration of specific tissues with age or injury and normal human movement. In a global sense, both disc degeneration and posture play a role in alteration of neural space (Panjabi et al., 1983). In testing of cadaveric specimens, Panjabi et al. (1983) demonstrated that spine flexion increases the size of the intervertebral foramen (IVF) by 24%, while extension decreases size by 20%. When cadaveric spines were tested with incremental loading in flexion-extension, lateral bending, and axial rotation, Nuckley *et al.* (2002) observed significant changes in IVF integrity during extension, lateral bending, and the combination of the two which mimicked physiological motions. A recent study of IVF measurements during dynamic activity (weight-lifting) also showed decreases in IVF area during flexion and extension movements (Zhong et al., 2015).

Nerve tissue compression in this region can specifically be due to facet hypertrophy, disc herniation, disc degeneration, disc height loss, or vertebral body height loss (Fontijne et al., 1992; Hashimoto et al., 1988; Nuckley et al., 2002; Teng, 1960). Overall, degenerated spine specimens have up to a 42% decrease in intervertebral foramen size compared to non-degenerated specimens (Panjabi et al., 1983). Modern research has aimed to quantify tissue strains during such conditions, but the interactions between these factors and mechanical thresholds remain unknown.

2.2.3 Compressive Loading, Fluid Loss and Disc Injury

The intervertebral disc consists of the nucleus pulposus, which acts like a pressurized fluid in healthy and slightly degenerated states (McNally & Adams, 1992; Nachemson, 1960, 1965), and the surrounding annulus fibrosus. The inner annulus has fluid-like behaviors despite its lamellar structure (Adams et al., 1993, 1994; McNally & Adams, 1992), while the fibrous outer annulus acts to resist tension and is the primary contributors during bending and torsional loads (Adams et al., 1994; Brinckmann & Grootenboer, 1991; McNally & Adams, 1992).

Upon compression, the nucleus distributes hydrostatic pressure across the vertebral endplates, which bulge into the vertebral bodies. The inner annulus also acts to resist compressive forces and bulge the outer annulus radially outwards (Brinckmann & Grootenboer, 1991; Stokes, 1988). The ability to distribute this hydrostatic pressure and resist load is directly linked to the fluid content and characteristics of the nucleus. During both acute compression and the cyclic compression exposures experienced over the course of a normal day, the disc loses water content and gains sodium and potassium (Kraemer et al., 1985). Further, decreased water content and nuclear volume in the disc have been associated with age, location, and disc degeneration (Antoniou et al., 1996; Gower & Pedrini, 1969). This altered water content is associated with changes in ultimate disc strength (Gunning et al., 2001) and decreased nuclear volume leads to decreased hydrostatic pressure and increased compression load resistance by the annulus (Adams et al., 1993). These content changes can also have detrimental effects on the annulus, as they make the annulus more elastic (Smeathers, 1984) and cause a reduction in resistance to bending (Adams & Dolan, 1996; Adams et al., 1987) and shear loads (Cyron & Hutton, 1981).

2.2.4 Axial Twist Loading and Tropism

During axial twist motions, fluid loss and spinal shrinkage are greater than in other motions (Au et al., 2001); however, the link between fluid loss and injury potential is not as straight-forward as during compression alone. Some loading exposures have shown facet damage with no structural damage to the IVD (Adams & Hutton, 1981), as facet joint contact occurs with as little as 1-3 degrees of twist in a healthy lumbar spine (Adams & Dolan, 1995). Conversely, similar load exposures have shown IVD damage without any boney (facet) damage (Farfan, 1969; Farfan et al., 1970), and annulus fibres resisted axial rotation more than the facet joints (Krismer et al., 1996) during combined compression and twist loading.

As with compressive loading and the intervertebral disc, facet injury risk can be significantly altered by not only posture and geometry modifiers, but also degenerative changes to the surrounding structures. Axial rotation of the spine compresses the articular surface of one of the two apophyseal joints. Under ideal conditions, upright twist-induced facet damage is unlikely since force is resisted across the broad, parallel joint surface. In full flexion, high stress concentrations may produce degenerative damages at the upper margins of the joint (Adams & Hutton, 1983). These postural differences can be exacerbated by changes to surrounding structures, as axial rotation motion is most affected by disc degeneration

(Fujiwara et al., 2000), allowing for greater range of motion and altered joint contact. Work with *in vitro* spine specimens has indicated the proportion of load can be greatly altered as the facet joints normally support about 20% of spinal compressive forces (Adams & Hutton, 1980; Adams et al., 1994), but can increase up to 70% with degenerative changes that narrow the discs (Adams & Hutton, 1980).

In addition to factors external to the facet joint and joint degeneration, internal joint geometry plays a large role in magnitude of vertebral rotation and overall joint strength. Facet joint tropism, or asymmetry between the left and right joint angles, can lead to greater magnitudes of rotational movement, and asymmetric load distribution. Tropism is generally measured relative to the bisection of the sagittal or frontal plane, and can be classified as no tropism (less than 5° or 6°), mild, moderate, or severe (greater than 15 or 16) tropism (Boden et al., 1996; Grogan et al., 1997). While most research indicates facet tropism does not influence the magnitude of rotation in cadaveric specimens (Gunzburg et al., 1991) and does not play a significant role in lumbar spine disc herniation (Ko & Park, 1997), there is some evidence that tropism can directly lead to disc degeneration (Adams & Hutton, 1983; Karacan et al., 2004). Further, *in vitro* annulus injury and nucleus removal resulted in asymmetric facet joint movement (Panjabi et al., 1984), indicating that intervertebral disc stability may play a role in facet tropism.

2.3 Axial Twist Load Modulators

2.3.1 Acute Twist

A combination of *in vitro* cadaveric studies and radiographs of *in vivo* human spines have reported maximum lumbar axial twist magnitude ranging from 2 to 5 degrees at each intervertebral joint (Adams & Hutton, 1981; Gunzburg et al., 1991; Pearcy & Tibrewal, 1984b) and up to 8 degrees for the whole lumbar spine (Rohlmann et al., 2001). A 3D radiographic technique demonstrated slightly more mobility at the L3-4 and L4-5 joints compared to the upper lumbar spine (Pearcy & Tibrewal, 1984b). Conversely, functional spinal unit failure has been shown to occur at maximum axial twist angles between 20.6 and

22.6 degrees (Drake, 2008; Farfan et al., 1970). The large difference between physiological range of motion and failure values can mostly be attributed to protocol, off-axis load, load rate and applied moment magnitude differences, but, most importantly, illustrate the effect loading parameters can have on modulating lumbar spine failure and movement in axial twist.

Expanding on previous work suggesting the primary function of the facet joints was to protect the disc from excessive torsion (Farfan & Sullivan, 1967), Farfan *et al*. (1970) evaluated overall torsion strength using cadaveric spines fixed in a gear-drive rotation assembly. Whole joint strength was approximately 88 N·m for specimens with normal discs and 54 N·m for specimens with degenerated discs. This load was subsequently partitioned into 14 N·m at the facet joint articular processes with an associated 14 degrees of rotation. These values were similar to later work that reported yield at 1-3 degrees of rotation when a 10-30 N·m moment was applied (Adams & Hutton, 1981). Farfan *et al*. (1970) performed a progressive dissection of the lumbar spine joint complex, and moment evaluation at each stage demonstrated approximately 90 percent of torsional load was resisted by the facet joints and the intervertebral disc, with an equal load share between those two respective elements. The authors suggested that torsional strength was dependent on the disc shape and load rate. In agreement with other studies, peripheral and circumferential annulus damage resembled changes that occurred due to degeneration, suggesting that annulus damage was a result of forced rotation, not compression (Adams & Hutton, 1981; Farfan & Sullivan, 1967; Farfan et al., 1970).

A more recent study investigated the role of static moment on functional spinal unit failure mechanics during highly repetitive flexion-extension motions with static compression (Drake et al., 2005). This study used C3-4 porcine spines to model the human lumbar spine as per several previous studies (Callaghan & McGill, 2001a; Oxland et al., 1991; Yingling et al., 1999). The addition of 5 N·m of static axial moment changed the failure mechanism relative to the flexion-extension loading protocol alone, as earlier onset of disc herniation, higher incidence of facet joint fracture, and higher energy dissipation were all observed

(Drake et al., 2005). While the applied axial moments in this study were modest relative to maximum moments between 12-32 N·m in other studies, the ability of a small magnitude axial moment to alter failure mechanics in combined loading illustrates the complexity of three-axis loading, and the need for future research to describe failure mechanics during physiological movements and loading rates.

2.3.2 Repetitive Twist

Evidence of failure tolerances during repetitive and/or fatiguing twist motions is limited, and most often difficult to compare due to differences in loading application method, load rate, load magnitude and off-axis load and posture. Liu *et al*. (1985) investigated the effect of cyclic torsional loads on the behavior of lumbar intervertebral joints. An initial protocol rotated specimens through 10 000 cycles at varying magnitudes of axial rotation and found a failure threshold of 1.5 degrees. Specimens rotated less than 1.5 degrees did not fail, while specimens rotated more than 1.5 degrees consistently failed in fewer than 10 000 cycles. Specimens were tested under a constant cyclic moment protocol and the resultant angular displacement was recorded, or under a constant angular displacement protocol and the resultant moment was recorded. Specimens with lower applied moments (11.3 and 22.6 N·m) resulted in angular displacements less than 1.5 degrees, while specimens with higher applied moments (33.9 and 45.2 N·m) exceeded 1.5 degrees threshold and failed in fatigue (Liu et al., 1985). Under displacement control, moment decreased from approximately ±10 N·m to ±4 N·m after 3000 cycles at ±1.5 degrees, indicating a creep response. Failure mechanism was very inconsistent between specimens, as failure modes consisted of end plate fracture, facet joint fracture, lamina fracture, and ligament damage.

Drake & Callaghan (2009) applied 1500 N of compression and repetitively loaded specimens in either flexion-extension or axial twist. The foramina pressure in the flexion-extension specimens was significantly higher than the axial twist specimens, and clear injury mechanism differences were shown. While all flexion-extension specimens herniated with 10 000 cycles of loading or less, 62.5% of twist specimens had incomplete herniation, 12.5% sustained facet fractures, and 25% had no evident damage (Drake & Callaghan, 2009). These findings act as further evidence to the complexity of injury mechanics during three-axis combining loading, and have implications on pain generation pathways. The increased foramina pressure during cyclic flexion-extension illustrates these motions as a viable nerve root compression pain generation pathway. Conversely, the differences during cyclic twisting agrees with other studies which have not yet yielded any consistent clinical predictors of pain resulting from occupational twisting motions (Adams, 2004; Schwarzer, Wang, Bogduk, et al., 1995).

2.3.3 Load Rate

Early research using gear-drive pulley systems (Farfan, 1969; Farfan et al., 1970) or rotating mounted plates (Adams & Hutton, 1981) generally used incremental loading of spine specimens and were unable to control for the rate at which the loads were applied. Such studies could not smoothly apply a ramped moment profile and were unable to immediately cease loading upon failure or identify other discrete changes in the slope of the moment-deformation curve (Drake, 2008). In agreement with expected physiological properties, limited recent research has shown lumbar spine tissues exhibit viscoelastic properties, meaning that load response is rate dependent. Yingling, Callaghan, & McGill, (1997) tested porcine C2-4 and C5-7 specimens to compression failure at five load rates (100, 1000, 3000, 10000 and 16 000 N/s). Dynamic loading increased the ultimate load and compression stiffness compared to the quasi-static load rate (100 N/s), and axial displacement at failure decreased with increased load rate. Further, load rate effected the failure mechanism, as low rate loading resulted exclusively in end plate fracture, whereas higher load rates lead to increased frequency of vertebral body fractures (Yingling et al., 1997). While some research has investigated rate dependencies of the isolated disc (Farfan et al., 1970; Race et al., 2000), few other studies have investigated the effect of load rate on failure mechanics of an *in vitro* model of the human lumbar spine and no research has evaluated the effect of load rate during twist motions.

2.3.4 Compression

When combined with repeated vertebral flexion-extension, compressive force as a secondary mode of loading has been shown to have a significant impact on the number of flexion-extension cycles to failure (Callaghan & McGill, 2001a; Parkinson & Callaghan, 2007b). Such compressive loads have a direct impact on facet joint articulation (Adams & Hutton, 1980), and stiffness changes due to increased compressive force may be due to changes in facet contact area and pressurization of the nucleus pulposus (Dunlop et al., 1984; Lin et al., 1978). Further, shear failure force is altered by facet joint articulation, as both postural deviation induced by shear load or disc height loss due to compressive load may alter moment arm characteristics of the facet joint centre of pressure (Howarth & Callaghan, 2012).

During axial twist, maximum compression tolerance is decreased with increasing twist moment magnitude, which may result from facet joint interface geometry changes (Aultman et al., 2004). The compression facet is the structure primarily responsible for resistance of axial twist load (Adams & Hutton, 1981). With increased compression, facet joint geometry and initial inter-articular facet joint spacing may be affected, which may lead to changes in the tracking and load carriage of both the compression and tension facets.

2.4 Porcine Cervical Spine Model and Specimen Considerations

Age-related changes, difficultly obtaining specimens, specimen variability and ethical considerations all make use of human specimens in spine injury research difficult. Young, healthy spines are typically preferred, but specimens from elderly and/or sick donors are most abundant for research use (Yingling et al., 1999). As such, porcine cervical models have been used as a surrogate for the human lumbar spine

(Callaghan & McGill, 2001a; Gunning et al., 2001; Howarth & Callaghan, 2012; Parkinson et al., 2005), and have demonstrated similar mechanical characteristics to a young adult with no disc degeneration or bone injury (Callaghan & McGill, 1995; Yingling et al., 1997, 1999). Several specimen considerations are discussed, and changes due to specimen state are outlined.

2.4.1 Storage

Due to cost-effectiveness, specimen transportation, testing duration, and multi-specimen requirements, frozen storage of biological materials for mechanical property testing is quite common. Frozen storage allows for large sample experiments on homogenous specimen groups, and these specimens can be frozen immediately after harvesting from the supplier in an attempt to maintain the physical state at the time of harvest. Several studies have evaluated the effects of freezing and thawing specimens, with results varying by methodology, tissue type and animal species (Callaghan & McGill, 1995). With respect to spine specimens, Callaghan & McGill (1995) found that frozen storage increased ultimate compressive load and energy absorbed to failure, but did not affect stiffness or displacement at failure. Conversely, Dhillon, Bass, & Lotz (2001) indicated frozen storage does not significantly alter the creep response in human lumbar discs, and noted that the subtle effects due to freezing do not alter the time-dependent response of the disc. Freezing and storage conditions have also shown no significant effect on specimen biomechanical properties (displacement due to anterior shear, axial rotation and lateral bending) when comparing fresh specimens to those frozen for a short or long duration (Panjabi et al., 1984). However, fresh specimens had greater variability in biomechanical properties than previously frozen specimens during a two week period (Panjabi et al., 1984). While specimens not loaded to failure seem to display minimal effects of frozen storage, storage medium is an important consideration for spine specimens loaded to failure.

2.4.2 Hydration

The nucleus of the intervertebral disc acts as a hydrostatic cushion during compressive loading, and partitions loads between the annulus fibrosus, facet joints, and other soft tissue structures. The hydrostatic pressure of the nucleus is a function of the fluid content, and the associated load distribution has a primary role in disc degeneration and herniation (Adams et al., 1996). Intervertebral disc fluid content varies with external load and load history, and when compressive stress exceeds the osmotic swelling pressure of the nucleus, fluid is expelled (Costi et al., 2002). When the disc is unloaded, it then absorbs fluid and returns to a steady state hydration level (Pflaster et al., 1997). Experimental methods attempt to maintain physiological hydration levels by considering storage techniques and active hydration methods during testing. Freezing and subsequent thawing for spine specimens results in increased swelling relative to fresh specimens, but preloading the specimen has been demonstrated to restore water content and disc load-deformation response to physiological levels (Adams & Dolan, 1996). The magnitude of this preload is important, as excessive preload may damage the specimen tissue, and specimen hydration has been shown to effect compressive failure tolerance (Gunning et al., 2001). During active testing, specimens have been immersed in a saline bath and showed a rapid rate of swelling during the first hour of immersion, followed by a decreased rate of swelling until a steady state was reached (Hirsch & Galante, 1967; Pflaster et al., 1997; Urban & Maroudas, 1981). Similar methodology has also indicated increased stiffness in a saline bath compared to specimens exposed to air (Costi et al., 2002). A group of studies from one biomechanics laboratory using a porcine cervical model demonstrated consistent load-deformation responses using a saline-soaked, plastic backed cloth. Callaghan & McGill (2001) showed no difference in final axial creep across a group of 26 specimens using a 260 N compressive preload. The same findings were presented by Aultman *et al*. (2004) and Drake *et al*. (2005) using a 300 N preload, which indicates that a 260-300 N preload may reliably normalize hydration levels across porcine specimens (Drake, 2008).

2.4.3 Specimen Age

Spine specimen age is often discussed in conjunction with degeneration, as degenerative changes often occur with age or injury. However, it is important to isolate the independent effects of age in healthy, uninjured specimens to speculate on degenerative changes. In a study using human cadaveric spines, Farfan *et al*. (1970) suggested that specimen tissues, specifically annular fibres, do not deteriorate purely as a result of age, but primarily due to damage and scarring. Also using adult lumbar spine specimens, Nachemson, Schultz, & Berkson (1979) demonstrated little effect of specimen age on mechanical properties. Specimen stiffness showed no significant effect of age, and intradiscal pressure changes in response to flexion or lateral bending were similar across ages. There was a minor effect on pressure changes in other movements, as increases were larger for older specimens in extension and shear, and larger for younger specimens in torsion (Nachemson et al., 1979). Pintar, Yoganandan, & Voo (1998) indicated an interaction effect between specimen age and loading rate, where young (30 year old) specimens had markedly more variability across loading rates than older (80 year old) specimens. Age effects were apparent as failure occurred at lower force values for older specimens when both old and young specimens were exposed to the same load rates (Pintar et al., 1998). However, it is difficult to distinguish these age-related differences from degeneration as a starting state for these specimens. These studies illustrate the importance of pre-screening and specimen grading prior to testing to isolate age-related changes from those resulting from pre-existing degeneration.

2.5 Summary

The purpose of the first *in vitro* study in this thesis was to validate the use of ultrasound as a measure of gross vertebral orientation during isolated trunk axial twist motions. The purpose of the second *in vitro* study was to investigate the role of axial twist magnitude and the associated effects on spine failure mechanics and injury tolerance. The existing published literature review provides justification for the methods used in these studies and overviews factors that should be considered in experimental design. The anatomy and underlying function of lumbar spine tissues were reviewed, and the effects of different factors that may alter tissue response to axial twist loading were discussed.

Passive testing has formed an understanding of normal lumbar spine anatomy and mechanical responses to one and two axes of cyclic and static loading. The geometric joint changes that occur with changes in water content, degeneration, and load distribution during compression and flexion-extension movements are well understood. Such structural and load distribution changes can alter the interactions between neural space and the mechanical thresholds for low back pain. However, during twisting motions, these mechanical and geometric changes are less clear. There remains disagreement about the role of the intervertebral disc during twist, and the combined effects of twist magnitude and load rate during physiological movements are unknown. Failure mode differences between flexion-extension and twisting protocols indicate that twist motions alter load distribution and changes during repetitive twist need to be investigated. To date, consistent clinical predictors of pain from occupational twisting do not exist and the effect of twist magnitude on cumulative load tolerance is unknown.

CHAPTER 3

GENERAL LITERATURE REVIEW FOR *IN VIVO* STUDIES

3.1 Overview

As discussed, two studies in this thesis used an *in vitro* porcine cervical spine model as a surrogate for the human lumbar spine, and the remaining two studies (Study 2 – Chapter 5 and Study 4 – Chapter 7) involved *in vivo* investigations of human motion and loading during twisting activities. With the nature of occupational twisting injuries discussed in Chapter 1 and lumbar spine anatomy, sources of pain, and movement mechanics discussed in Chapter 2, this chapter outlines the physical demands and reach capacities of *in vivo* participants during simulation of occupational tasks in a laboratory environment. Reach envelope development is discussed with reference to the effects of hand location and exertion type on reach and strength capacity. Specific details and motivation for the *in vivo* studies in this thesis are included in the background and introductory information provided with the appropriate chapter.

3.2 Muscular Co-Activation during Occupational Twisting

Occupational axial twisting is a prominent risk factor for low back disorders, yet the mechanism for muscular generation of twisting moments during twisting alone and coupled movements are not fully understood. Investigations of muscle activation estimates from surface electromyography have described significant trunk muscle co-activation during twisting movements. In a study of muscle activation during controlled isometric axial torques, Pope *et al*. (1986) indicated the highest activation in internal and external obliques on the side of rotation development, but also found substantial antagonistic activity in muscles not directly involved in development of the motion-inducing moment including erector spinae. This co-activation appeared to be a function of the coupled flexion-extension posture. When the trunk is pre-rotated to one side by 30 degrees, the ability to produce torque decreased when rotated towards the direction of rotation and increased when rotated away from rotation (Pope et al., 1987). McGill (1991) observed trunk muscle activity during isometric and isokinetic twisting exertions. During maximal isometric twist activities, maximum activation was 52% MVC for external oblique, 55% MVC for internal oblique, 61% for upper (T9 level) erector spinae, and 33% MVC for lower (L3 level) erector spinae. The authors suggested balancing of flexion-extension and lateral bending moments as a primary contributor to these less than maximal activation levels. The levels of erector spinae activation suggest significant co-contraction involved in these exertions. Further, isokinetic exertions elicited high levels of erector spinae activation despite no mechanical potential to contribute to the axial twist moment (McGill, 1991).

Under various force, velocity, position and direction conditions, Marras & Granata (1995) assessed activation for 10 trunk muscles and modelled joint moments. Significant flexion-extension and lateral bending moments were present during twisting exertions, and showed varied spine compression and lateral shear load depending on the amount of twist when rotation was applied and the direction of the applied rotation. Similar to other studies, co-activation was high during twisting exertions, and was over 60% greater than co-activation levels observed during lifting exertions. To further characterize muscle activation patterns during coupled twisting movements, Marras *et al*. (1998) investigated twisting movements during upright standing as well as from flexed and asymmetric postures. From standing, agonist muscular demands were relatively low, as the obliques and latissimus dorsi assisted in development of the rotational torque. From flexed and asymmetrically bent postures, erector spinae activity was increased by up to 85% and external oblique activity decreased by up to 20%. These findings indicate that muscle demands and contribution to rotational moment change as a function of lumbar spine posture. There remains a need to investigate muscular demand changes during *in vivo* occupational twisting motions, and the relationship between muscular demands, joint loading, and work or hand target location needs to be described. Such a relationship will assist in establishing relevant workstation reach guidelines to mitigate muscular demands.

3.3 Reach Envelopes

Nearly all industrial tasks involve some degree of arm reach to retrieve various tools or parts, or to operate hand controls at a workstation. Several investigations have outlined maximum workspaces or reach envelopes in various jobs types that identify dimensions for workers with limiting (5th percentile) reaches. Kennedy (1978) compiled a three-dimensional reach capability data set targeted for aircraft design. Participants completed a series of reaching tasks to develop a two-dimensional (forward-lateral) reach envelope at various vertical planes. Stoudt *et al*. (1970) performed a similar data collection to establish locations of automotive controls with respect to reaching limits. Both these studies developed cylindrical reach zones, which were not related to abdominal depth or elbow height. Haslegrave (1986) demonstrated large amounts of variation between limbs lengths within a given stature, and suggested error would occur if the 5th percentile maximum reach envelopes for industrial workstations were constructed additively using 5th percentile values of several anthropometric dimensions. It has also been suggested that the tilt and backrest angle of an automotive or airline seat would induce error compared to reaching capabilities in a normal industrial seat (Sengupta & Das, 2000).

For industrial workstations, Faulkner & Day (1970) directly measured seated reach distances for female workers, and Behara (1991) measured both male and female 5th percentile worker seated reaches for various North American industrial occupations. Each of these investigations involved measurement of reach at a set of uniformly distributed angular directions, and did not account for occupational factors such as reaching effort, trunk posture, shoulder posture, or standing postures. Sengupta & Das (2000) measured dynamic reach envelopes for 5th, 50th and 95th percentile male and female participants during both standing and seated work positions using a digital potentiometer system. Reach was significantly greater while standing than while sitting, and greater for males. Thus, guidelines for standing work and sitting work may need to be different. Current ergonomic design guidelines for Canadian workplaces (CSA Z1104-12) provide forward and lateral reach dimensions in primary, secondary, and tertiary reach zones

(Figure 3.1). These zones relate to “frequent”, “infrequent”, and “occasional” reaching tasks, respectively. While these guidelines and described studies provide detailed reach capabilities related to worker anthropometry, they do not necessarily consider physical demands during such reaches.

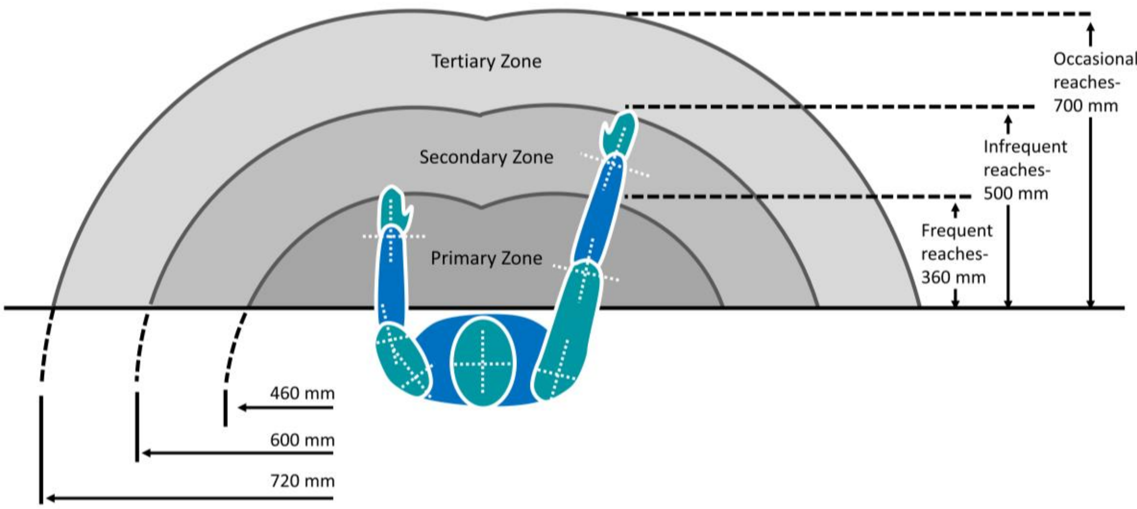


Figure 3.1: Current ergonomic reach guidelines. Primary, secondary and tertiary reach zones are intended for “frequent”, “infrequent”, and “occasional” tasks as described in current guidelines. Figure adapted from CSA (2012).

The effect of task location on physical demands and perceived discomfort have been studied with a primary focus on the upper limb. Many investigations have been performed to determine task- or posture-specific strength capacities. Fewster & Potvin (2014) evaluated maximum in-line push strength for males and females and identified 25th percentile strength values for use in digital human analysis and ergonomics guidelines. La Delfa *et al*. (2014) developed a regression model to predict manual arm strength based on 28 hand locations within a reach envelope. The regression equation explained 92.5% of the variance using anterior, lateral, and vertical location of the hand relative to the active shoulder as inputs. McDonald *et al*. (2014) and McDonald *et al*. (2012) developed a three-dimensional spatial map of the right upper limb during submaximal lateral and push-pull exertions, respectively. Rightward (away from midline) exertions showed the highest total muscle demand at 20 cm laterally, 50 cm forward and 20 cm below an origin centred between the hips at the level of the umbilicus. Leftward exertions presented greater demands when moving leftwards across the midline, and showed similar trends to rightward exertions for forward and vertical distances (McDonald et al., 2014). Pushing and pulling exertions, in general, elicited increased demands moving laterally from the midline, moving closer than 30 cm towards the body, and moving above the level of the origin (McDonald et al., 2012). These studies demonstrate upper limb strength capabilities related to hand location, but most often do not describe the associated low back kinematics and associated physical demands. More specifically, investigations are required to link lumbar segmental motion to torso and upper limb kinematics during a range of forward and lateral reaches.

3.4 Summary

The purpose of the first *in vivo* study in this thesis was to develop a relationship between segmental vertebral axial twist and external thoracopelvic twist. The purpose of the second *in vivo* study was to investigate external axial twist during simulated occupational tasks across a range of forward and lateral reach distances and task characteristics. Studies of manual material handling and twisting exertions have identified high levels of trunk muscle co-activation during controlled, isometric twisting exertions, and show a 60% greater level of co-activation than occupational lifting tasks. These investigations describe off-axis moment development as a primary reason for co-activation, but cannot yet define the relationship between muscular demands, joint loading, and work or hand target location during occupational twisting motions.

Upper limb investigations have defined the relationship between three-dimensional hand location and upper limb strength. These studies have both assisted in and progressed from the development of workplace reach envelope guidelines. Such guidelines are necessary for ergonomic design to reduce or mitigate physical demands on the low back and upper limb during reaching tasks. Current guidelines are mostly driven by anthropometric measures, and need to be revised to include the low back posture changes and associated physical demands as a function of target location, exertion direction, and hand load characteristics.

CHAPTER 4

STUDY #1: VALIDATION OF AN ULTRASOUND PROTOCOL TO MEASURE INTERVERTEBRAL AXIAL TWIST DURING FUNCTIONAL TWISTING MOVEMENTS IN ISOLATED FUNCTIONAL SPINAL UNITS

4.1 Introduction

Epidemiological and biomechanical studies have identified specific occupational risk factors associated with low back disorders (Marras, 2000; Marras et al., 1993) such as static postures (Kelsey, 1975; Magora, 1970, 1973), frequent bending and twisting (Andersson, 1981; Kelsey et al., 1984; Keyserling et al., 1988), manual exertions (Chaffin & Park, 1973; Kelsey & White, 1980; Kelsey, 1975; Kelsey et al., 1984), and repetitive movements (Kelsey et al., 1984; Rubin et al., 1987). Developing a link between these risk factors and specific spine joint postures is important for understanding age-related and work-related changes in spine integrity.

In order to relate occupational low back disorder risk factors to joint posture changes in the lumbar spine, a strategy for task-orientated investigations of intervertebral motion is required. External measurement of intervertebral motion has been performed using markers attached to the skin (Lee et al., 1995; Mörl & Blickhan, 2006), but skin motion artifact may cause differences between surface measures and actual intervertebral motion (Heneghan & Balanos, 2010). Tracking of markers attached directly to the vertebral spinous processes (eg. bone pins) may eliminate this error (Steffen et al., 1997), but is extremely invasive and impractical for task-oriented twisting investigations.

Intervertebral motion can be accurately measured using medical imaging modalities. Studies measuring axial twist using magnetic resonance imaging, computed tomography, and radiography (Haughton et al., 2002; Ochia et al., 2006; Pearcy & Tibrewal, 1984b) have each been able to identify segmental rotations, but demonstrated inconsistencies across vertebral levels between studies. Both Haughton *et al*. (2002) and Pearcy & Tibrewal (1984b) indicated increased segmental rotation inferiorly from L1-L2 to L4-L5. These findings correspond with the use of L4-L5 as the primary site of movement in many modelling approaches (Callaghan & McGill, 2001b; Cholewicki et al., 1991). Conversely, Ochia *et al*. (2006) reported lower axial rotation at L4-5 than any other lumbar level, and suggested that facet joint orientation differs at each level resulting in range of motion differences.

It is difficult to assess mechanisms for differences between vertebral levels, as these studies did not evaluate the same external rotations, and differing flexion-extension postures may have invoked vertebral joint geometry alignment changes. Further, magnetic resonance imaging and computed tomography present some challenges for task-oriented upright postures and dynamic investigations, as they require a participant to be in a fixed posture within a confined imaging space.

In contrast, ultrasound does not require a patient or participant to be secured to an imaging device, and can be used during occupational, task-specific movements. Ultrasound has been used and validated in the lumbar spine to quantify sagittal plane posture (van den Hoorn et al., 2016) and in the cervical spine to identify boney structures and adjacent nervous tissues (Martinoli et al., 2002; van Eerd et al., 2014). Other studies have used ultrasound as a guidance technique for local injections in the lumbar facet joints (Galiano et al., 2005, 2006, 2007). It is theorized that similar techniques to these lumbar and cervical studies can be employed to evaluate boney movement during axial twist of the lumbar spine in both *in vivo* and *in vitro* evaluations. Segmental rotations could then be measured under controlled external thoracic axial twist conditions and in response to mechanical loading.

The purpose of this study was to measure vertebral segmental rotations in a porcine model of the human lumbar spine using an ultrasound imaging protocol, and to validate use of this imaging technique with an optical motion capture system.

4.2 Methods

4.2.1 Specimen Preparation

Twelve fresh-frozen porcine spines (C1-T1) were obtained from a common source to ensure similar physical activity, diet, and age between specimens prior to death. Each specimen was thawed overnight prior to testing at room temperature, and dissected to an osteoligamentous functional spinal units (FSU; two vertebrae and the intervertebral disc) (equal number of C3-C4 and C5-C6). Each specimen was mounted into custom machined aluminum cups and secured using wire wrapped around the laminae and anterior processes, and a fixation rod inserted transversely through the vertebral body (Figure 4.1A). Two imaging pins were inserted into each vertebral lamina to facilitate sonogram point tracking (Figure 4.1B). FSUs were then surrounded by a water-filled imaging tank to allow ultrasound imaging of the FSU during experimental trials (Figure 4.2).

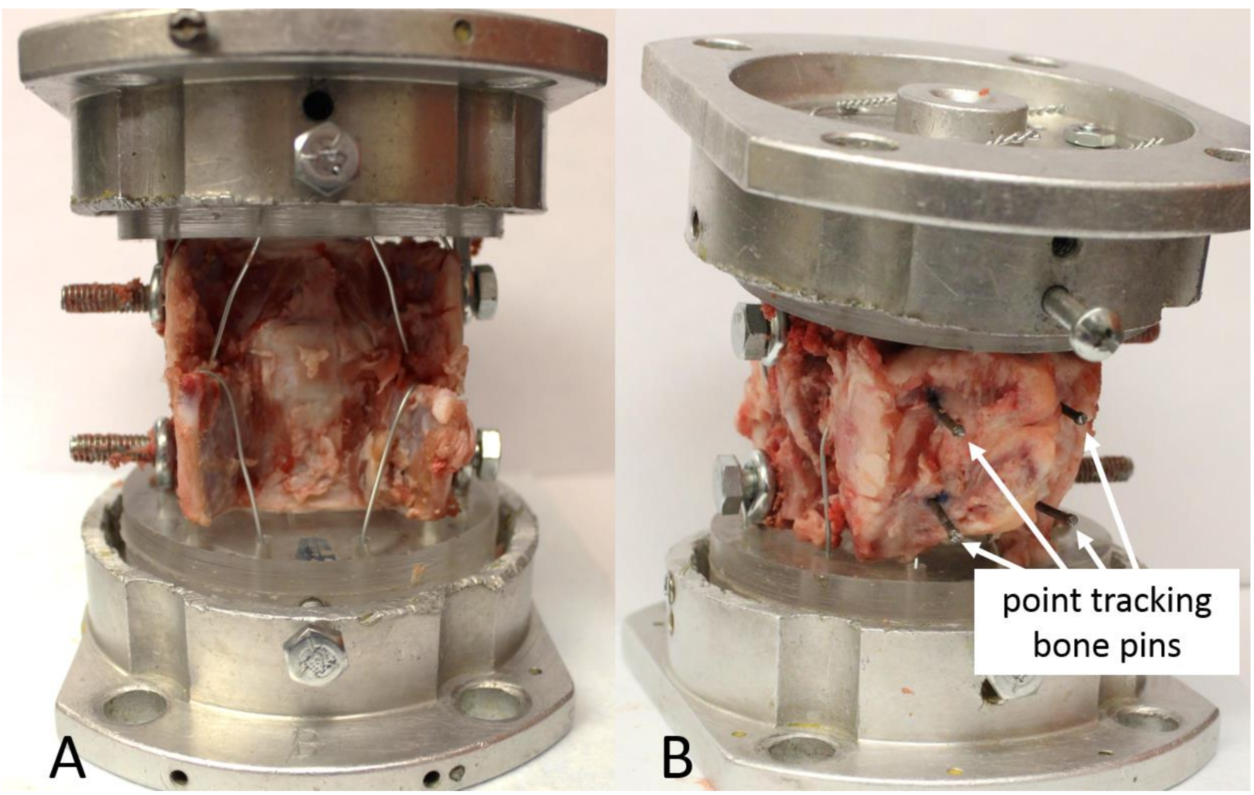


Figure 4.1: Spine specimen mounted in aluminum cups for testing in the servo-hydraulic dynamic testing system. A) Specimen is mounted using wire and lateral fixation rod (anterior view); B) posterior laminar imaging pins for sonogram point tracking (posterior view).

Similar setups for vertebral ultrasound imaging have been employed for both cervical (van Eerd et al., 2014) and lumbar spine specimens (van den Hoorn et al., 2016). The specimen and imaging tank assembly was mounted to an aluminum block base and inserted into a modified servo-hydraulic dynamic testing system (Model 8872, Instron Canada, Burlington, ON, Canada) that allowed load-control in the compressive axis, and position-control in the flexion-extension (FE) and axial twist (AT) axes. FE position was controlled using a servo-controlled torque motor and custom-built specimen carriage, and AT angle was controlled using a steel rod threaded through the aluminum block base and a pair of servo-controlled linear actuators (RSA24, Tolomatic Inc., Hamel, MN, USA) driven by brushless servomotors (AKM22E, Danaher Motion Inc., Radford, VA, USA; Figure 4.2). Actuators moved in opposite directions to create a force couple about the centre of the FSU.

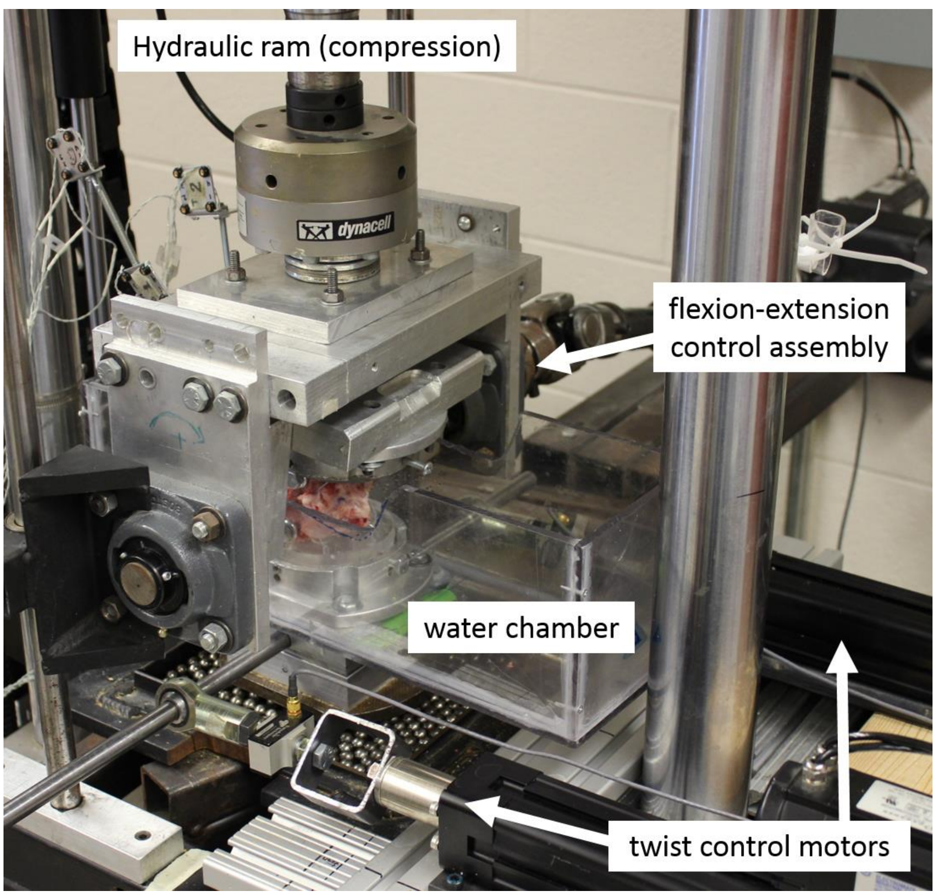


Figure 4.2: Spine specimen mounted in loading apparatus and surrounded by a water chamber. Flexion-extension motion is controlled by a custom control assembly, and axial twist motion is controlled by a pair of linear actuators.

4.2.2 Loading Protocol

Each specimen was preloaded with a 300 N compressive load at zero flexion/extension moment for 15 minutes to counter any post-mortem intervertebral disc swelling (Callaghan & McGill, 2001a). The angular position at the end of the preload was used as the zero position for further testing of each specimen (Drake et al., 2005). Specimens were then loaded in a minimum of five repeats of a flexion-extension range of motion test at a rate of 0.5 degrees per second (Callaghan & McGill, 2001a). The points at which resistance to motion (when the moment versus angular position curve deviates from the initial linear section) initiated were defined as the static flexion and extension limit angles used for experimental trials. This measure mimics the neutral zone defined by Panjabi *et al*. (1989).

For experimental trials, specimens were loaded in three axes. A compressive load was applied to the FSU specimen during all trials, which corresponded to 15% of the predicted ultimate compressive strength tolerance as calculated from a linear regression calculation based on endplate area (Parkinson et al., 2005). Global system axial twist was position-controlled to achieve one of four levels (0°, 2°, 4°, and 6°) of leftward axial twist from the zero position. These twist magnitudes fall within the range observed in the human lumbar spine with less than 7.5 N·m torsional moment (Rohlmann et al., 2001). These axial twist magnitudes were performed with the specimen in three sagittal plane postures (neutral, flexion, and extension) for a total of 12 experimental trial conditions.

4.2.3 Vertebral Axial Twist Measurement

Threaded rigid body fixation screws were inserted into the anterior side of the superior and inferior vertebral bodies and extended vertically to exit the imaging tank. A rigid body containing four infrared markers was attached to the terminal end of each fixation screw and marker positions were recorded during each trial using an optical motion capture system (Optotrak Certus, Northern Digital Inc., Waterloo, ON) at 32 Hz. Posterior, transverse plane sonograms were recorded using a 13-6 MHz linear ultrasound transducer and a medical ultrasound system (M-Turbo, Sonosite, Bothel, WA). Separate sonograms were recorded for each vertebral level (superior and inferior) in which both laminar imaging pins were visible. Penetration depth, resolution, and intensity were kept constant for all sonograms within each specimen. A rigid body containing four infrared markers was also mounted to the ultrasound transducer handle. Virtual markers were digitized at several locations on the ultrasound transducer. These virtual markers were tracked based on a fixed rotational and translational relationship with the rigid body.

4.2.4 Vertebral Angle Definition

Vertebral axial twist was characterized using two methods: 1) relative intervertebral angle of the vertebral marker rigid bodies about the vertical axis of the inferior vertebra, and 2) relative intervertebral laminar imaging pin angles (from ultrasound images) about the vertical axis of the inferior vertebra. All motion analysis was performed using MATLAB (2015a, Natick, MA, USA)

Method 1 (vertebral marker rigid bodies – motion capture, MC): A local coordinate system (LCS) was developed for the inferior vertebra (V_INF) using International Society of Biomechanics (ISB) convention for spine motion (Wu et al., 2002). LCSs were also defined for marker rigid bodies (MC_SUP, MC_INF) using a similar convention (anteroposterior x-axis, vertical y-axis, mediolateral z-axis). Three-axis rotation angles between the bone pin local systems and V_INF were calculated using a YZX Euler rotation sequence.

Method 2 (laminar imaging pins – ultrasound, US): For the ultrasound transducer, a LCS was created to define the sonogram location in global space (z-axis: along transducer face; y-axis: along handle length; x-axis: perpendicular to y- and z-axis). Laminar imaging pin positions were identified in the transducer local system using a custom-written image analysis program (Figure 4.3), and positions were transformed into the global system axes. Imaging pin positions were digitized three times by the same experimenter in order to calculate intra-rater error. A vector between the two laminar imaging pin points, using the mean of the three digitizations, was calculated for both the superior (US_SUP) and inferior vertebrae (US_INF), and three-axis rotation angles of this vector about V_INF were calculated using a YZX Euler rotation sequence.

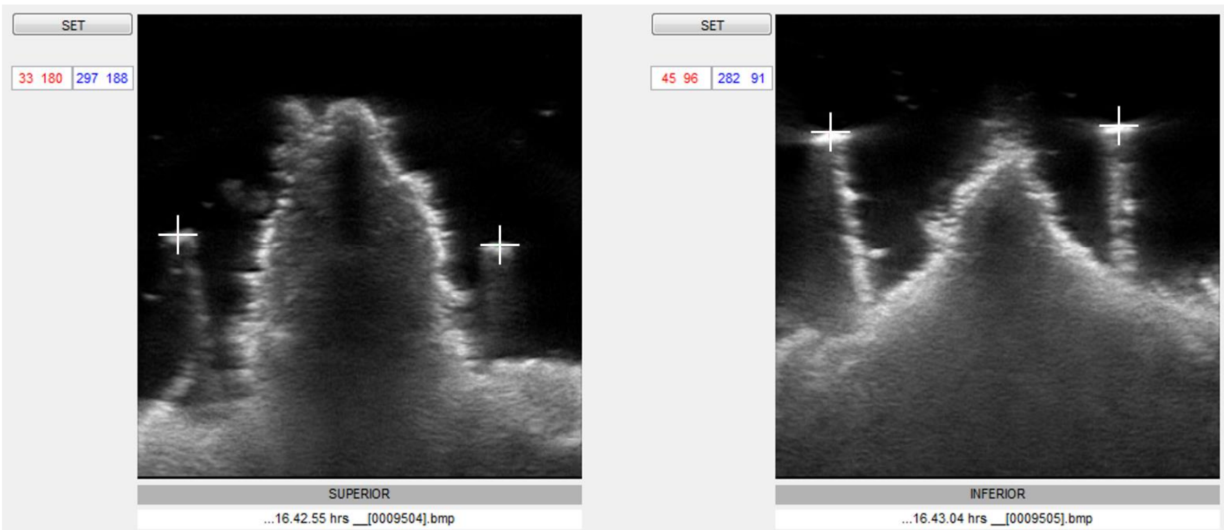


Figure 4.3: Example of laminar imaging pin position identification using a custom-written image analysis application.

4.2.5 Data Reduction and Statistical Analysis

All rotation angles were normalized to the zero position and leftward axial twist angle changes are expressed as absolute deviation from the zero position, in degrees. For both the MC and US measurement protocols, angles are expressed as the axial twist difference between the tracking system measurements of the superior and inferior vertebrae about the V_INF vertical axis.

To evaluate the relationship between the intervertebral angles measured using ultrasound and motion capture, a linear regression between the two systems was performed, and correlation (Pearson r) was calculated. Absolute error between methods was also calculated. To evaluate the effect of system on intervertebral angle across FE postures and AT magnitudes, a general linear model (measurement system; FE posture; applied axial twist angle) with Tukey HSD post-hoc analysis and Bonferoni correction was used with statistical significance set at $\alpha = 0.05$. All statistical analyses were performed using R-Studio (v1.0.136, Boston, MA, USA). All results are collapsed across C3-C4 and C5-C6 specimens and are presented as mean ± standard error, unless otherwise indicated.

4.3 Results

Mean flexion and extension postures across all specimens were 5.58 ± 1.2 degrees and 2.37 ± 1.7 degrees from neutral, respectively. Correlation between the two measurement systems showed similar fits regardless of the measurement approach (Figure 4.4). Pearson product-moment correlation (r) between the two systems was 0.903 overall (neutral: 0.897; flexion: 0.901; extension: 0.907). Slopes and y-intercepts for the linear fits are presented in Figure 4.4. Absolute system error was 0.014 ± 0.05 degrees overall, -0.060 ± 0.09 degrees in a neutral flexion-extension posture, 0.008 ± 0.10 degrees in flexion, and 0.094 ± 0.08 degrees in extension. No significant effects were observed for any interaction (p=0.87 to 0.98) or main effect (p=0.85) involving *measurement system*. A significant *applied angle*-by-*FE posture* interaction effect (p=0.046) and an *applied angle* main effect (p<.0001) were observed for intervertebral axial twist angle. Intervertebral angle increased with increasing applied angle, and this angle change was greater for specimens in a flexed posture than in neutral (p=0.01) or extension (p<.0001). Mean intra-rater error across three iterations of imaging pin digitization was -0.050 ± 0.009 mm in the anterior posterior direction and -0.024 ± 0.007 mm in the mediolateral direction.

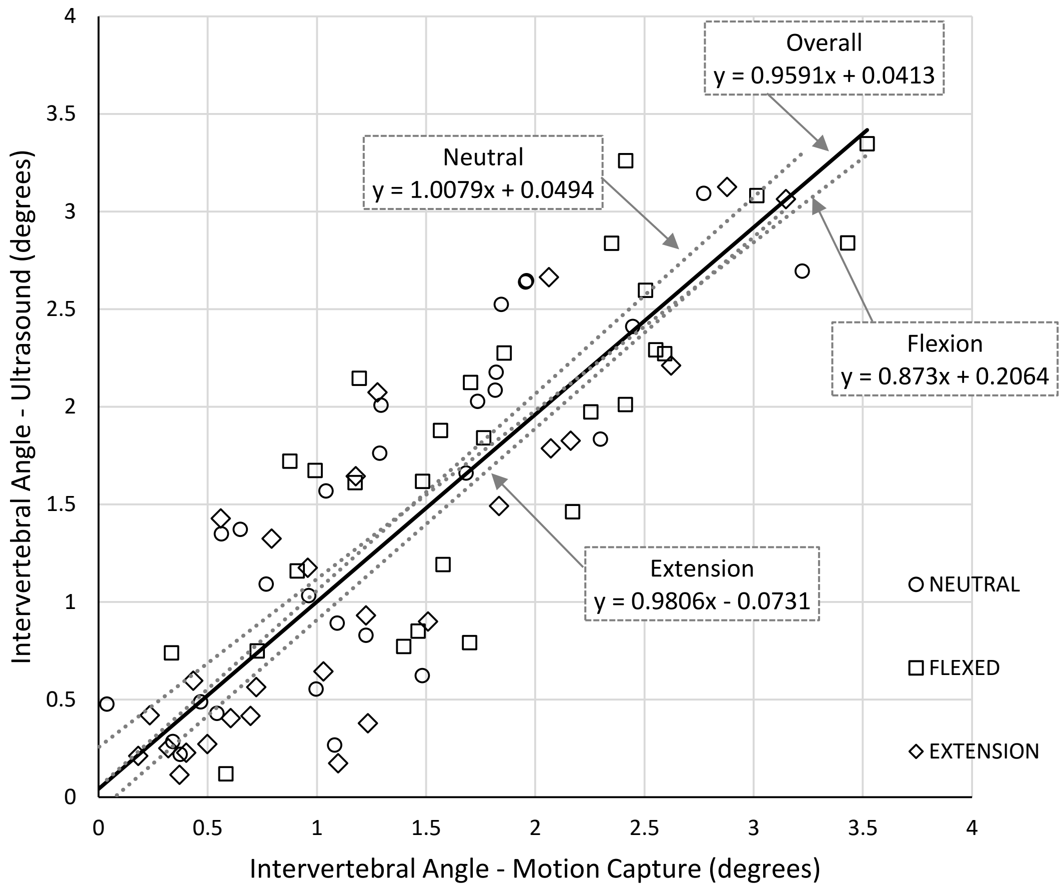


Figure 4.4: Intervertebral angle measured using a motion capture system (x-axis) and ultrasound system (y-axis). Deviations from the linear fits represents error between the two systems.

4.4 Discussion

The results demonstrate that vertebral axial twist can be accurately measured using ultrasound to track landmarks fixed to boney geometry. This measurement showed very low error compared to vertebral axial twist measurements from an optical motion capture system, which can be considered the Gold Standard, in this experiment, for spatial motion.

The average error between the two measurement systems was less than 0.1 degrees across the evaluated range of motion, regardless of flexion-extension posture, with the greatest error in an extended posture (0.094 ± 0.08 degrees). These errors are lower than an investigation using ultrasound to track sagittal plane lumbosacral motion (2.1 degrees) (van den Hoorn et al., 2016). However, the maximum intervertebral motion in the current study (3.5 degrees) was lower than that study (14 degrees), and may contribute to these error differences. The tested range of motion differences are justified, as intervertebral axial twist motion is generally less than 3 degrees (Haughton et al., 2002; Ochia et al., 2006; Pearcy & Tibrewal, 1984a), whereas lumbosacral flexion-extension range of motion is nearly 14 degrees (White & Panjabi, 1978).

Inter-system correlation had similar values across each of the three evaluated flexion-extension postures (neutral, flexed, extended), and all postures had a correlation of 0.9 or greater. The success of the ultrasound technique between flexion-extension postures is critical for potential success for use during *in vivo* measurement of axial twist as, unless specifically controlled, human axial twist motion is typically coupled with some degree of flexion-extension motion.

The overall y-intercept and slope of the linear fit to the data scatter were 0.04 and 0.96, respectively, which indicates that the two measurement systems have 0.04 degrees of error at baseline, and ultrasound tends to slightly underestimate motion capture measures of intervertebral angle with increasing angle magnitude.

The sonogram quality was very high due to the absence of any soft tissue and the ability to rigidly affix imaging pins to the bone. Imaging pins were not affected by submersion in water, and provided a high level of intra-rater consistency for sonogram landmark digitization. The mean anterior-posterior error in multiple digitizations of the same image (0.05 mm) across the mean width between the imaging pins (226.9 mm) would equate to a 0.001 degree angular error for the vector generated from the two digitized imaging pin points.

The same point digitization process on *in vivo* specimens will certainly present more difficulty than in the current study. The small amount of soft tissue remaining on the dissected osteoligamentous specimens in the current study became saturated and sonographically opaque when submersed in the water-filled imaging tank. This saturation made consistent identification of boney margins difficult, and required digitization of imaging pins rather than bone geometry positions. Tracking boney landmark margins (without the ability to use imaging pins) will be similarly challenging during *in vivo* measurement of vertebral axial twist, and future investigations must develop a strategy for consistent point identification and relative distance measurement between digitized points.

The *applied* axial twist angle did not match the *measured* intervertebral angles due to uncoupling of the specimens from the mounting apparatus. The facet joints protect against axial twist moment and resist twist motion (Adams & Hutton, 1981; Pearcy & Hindle, 1991), and this high resistance to rotation transferred twist moment from the intervertebral joint to the mounting surface and caused motion between the specimen and the mounting apparatus – typically at the superior side. This uncoupling does not affect the comparison of the ultrasound and motion capture methods for measuring axial twist, but reduced the maximum evaluated range of motion from potentially 6 degrees (maximum applied twist magnitude) to 3.5 degrees (maximum measured intervertebral angle). Uncoupling was greatest in extension (71.8% of applied range of motion), followed by neutral (63.8%) and flexion (48.8%). The significant *applied angle*-by-*FE posture* interaction effect can be attributed directly to the uncoupling difference by FE posture. While the applied axial angle was identical for all FE postures, uncoupling was lowest in flexion resulting in a greater intervertebral angle. With a natural extension (lordosis) in the FSU, flexion set the specimen closest to a vertically stacked vertebral posture. This posture created more adequate interaction with the plaster and wire fixation apparatus and reduced the overall uncoupling.

This investigation indicates that ultrasound can be used to track axial twist motion in an *in vitro* spine motion segment with low error and high correlation with a motion capture system. This technique has the potential for use *in vivo* to evaluate absolute intervertebral motion about the axial twist axis, as well as to determine how motion differs across vertebral levels and during coupled motions. Having estimates of internal vertebral geometry changes associated with gross external thoracopelvic motion may help investigate joint posture linkages to low back axial twist injury mechanisms and working posture guidelines.

4.5 Conclusion

Intervertebral axial twist measurement agreement between an ultrasound measurement technique and a motion capture technique (considered Gold Standard) showed high correlation across a physiological range of axial twist postures and between neutral, flexed and extended postures. The high level of agreement between the systems promotes an investigation of its *in vivo* use to link external thoracopelvic motions to segmental vertebral motion during simulated working tasks in an effort to link spine posture changes to occupational risk factors and injury outcomes.

CHAPTER 5

STUDY 2: THE RELATIONSHIP BETWEEN EXTERNAL THORACOPELVIC ANGLE AND LUMBAR SEGMENTAL AXIAL TWIST ANGLE USING AN ULTRASOUND IMAGING TECHNIQUE

5.1 Introduction

Decades of research has sought to isolate specific biomechanical aspects of work that contribute to a continuing rise in worker pain and injury in the modern work era, despite increased automation and reduced workplace physical demands. Review studies are effective in identifying correlations between physical work and injury prevalence. These studies typically identify heavy loads, lifting and forceful movements, whole body vibration, static postures, and bending and twisting as factors that will likely lead to low back disorders (Bernard, 1997; Hoogendoorn et al., 1999; Marras, 2000). The latter, bending and twisting, are required postures and occupational demands in both classic industrial (Marras et al., 1993) and modern mobile work environments (McKinnon et al., 2011). Biomechanical studies have presented similar findings when evaluating spine motion and loading. Marras *et al*. (1993) and Marras *et al*. (1995) quantified physical exposures at the lumbar spine in more than 400 jobs and identified lifting frequency, sagittal torso flexion angle, lateral velocity, twisting velocity, and external load moment as key factors in predicting risk of low back disorder. Other *in vivo* evidence indicates that combining lifting and twisting increased injury risk for workers whose jobs involved those tasks (Kelsey et al., 1984). While the risk related to occupational twist is apparent, the joint-level changes that occur in the lumbar spine are not well understood and difficult to assess with surface kinematic measures.

Lumbar spine load and motion analyses classically use the L4-L5 intervertebral joint as the primary site of motion and to represent the low back exposure (Callaghan & McGill, 2001b; Cholewicki et al., 1991). While appropriate for many modelling scenarios, this simplification does not necessarily reflect the motion and loading that occur at individual vertebral joints during a task. During *in vivo* evaluations using radiographs (Pearcy & Tibrewal, 1984b) and Steinmann pins (Gunzburg et al., 1991) maximum lumbar axial twist angles were approximately 2 degrees between vertebral levels. In comparison, *in vitro* torsion limit testing of cadaveric human spines showed intervertebral angles between 2 and 5 degrees (Adams & Hutton, 1981; Gunzburg et al., 1991) and aggregate twist of up to 8 degrees across the entire lumbar spine (Rohlmann et al., 2001). While radiographs of the lumbar spine have presented slightly more mobility at the L3-L4 and L4-L5 joints compared to the upper lumbar spine (Pearcy & Tibrewal, 1984b), an MRI investigation suggested that more trunk rotation takes place in the thoracic spine than the lumbar spine (Fujii et al., 2007). The latter does not support the use of a single site of rotation (L4-L5) in the lumbar spine, but rather that thoracopelvic motion is distributed across the thoracic and lumbar intervertebral joints.

In flexion-extension, when moving from standing to sitting postures in males intervertebral joint angles have been shown to differ based on level, with a disc angle of approximately 5 degrees at the L1-L2 level and increasing up to 14 degrees at the L5-S1 level with a 40 degree aggregate lumbosacral angle, or lumbar angle (De Carvalho et al., 2010). Such radiographic measurements of lumbar angle have been suggested as the most valid and accurate measures of lumbar lordosis (De Carvalho et al., 2010; Polly et al., 1996), but very few studies have partitioned lumbar angle to individual segmental motions.

In axial twist, segmental motion has been quantified using radiographs (Pearcy & Tibrewal, 1984b), computed tomography (Ochia et al., 2006), magnetic resonance imaging (Haughton et al., 2002), and *in vitro* methods (Panjabi, Yamamoto, et al., 1989). Some of these studies reported increased segmental rotation from L1-L2 to L4-L5 (Haughton et al., 2002; Pearcy & Tibrewal, 1984b), while others showed greater axial rotation in the upper lumbar spine (L1-L2 to L3-L4) compared to L4-L5 and L5-S1. Interestingly, Ochia *et al*. (2006) indicated the lowest axial rotation magnitude at L4-L5, which is typically used as the representative site of axial rotation for the torso as a whole. Ochia *et al*. (2006) suggested that segmental motion at different levels may be the result of facet joint orientation differences (facets oriented more sagittally from L1-L2 to L4-L5 and more coronally in L5-S1). The gross external rotations were not consistent across all of these previously mentioned imaging studies, and flexion-extension postures were not reported or controlled. These factors may explain the differences between the intervertebral angles between studies since axial twist range of motion is altered by flexion-extension posture (Drake, 2008; Gunzburg et al., 1991).

Computed tomography and magnetic resonance imaging are effective and non-invasive methods to quantify segmental twist motion; however, both systems present some limitations in recording axial rotation movements *in vivo* and have limited use as a field modality. Both modalities require a patient to be secured to an imaging device, which limits the ability to perform occupational movements and restricts the use of external fixation to control secondary and tertiary axis movements in coupled postures. In comparison, ultrasound is a relatively inexpensive imaging modality and is more broadly available to research labs than MRI or CT. Ultrasound has been used and validated in the lumbar spine to quantify sagittal plane posture (van den Hoorn et al., 2016) and in the cervical spine to identify boney structures and adjacent nervous tissues (Martinoli et al., 2002; van Eerd et al., 2014). Other studies have used ultrasound as a guidance technique for local injections in the lumbar facet joints (Galiano et al., 2005, 2006, 2007) and for measurement of shear vertebral displacement (Kawchuk, Fauvel, et al., 2001; Kawchuk, Kaigle, et al., 2001), but has not yet been used as a modality to assess segmental motions during gross axial twist movements.

Therefore, the purpose of this study was to evaluate lumbar segmental axial twist in relation to external thoracopelvic twist using an ultrasound imaging technique. This study will allow for conclusions about how each vertebral level contributes to lumbar spine motion during axial twist movements and give potential for evaluation of physiological levels of axial twist using *in vitro* techniques to evaluate injury potential during twisting activities commonly seen in industrial environments.

5.2 Methods

5.2.1 Participants

Sixteen (8 male, 8 female) participants aged 19-29 were recruited for this study (mean ± standard deviation – age: 24.8 ± 3.8 years; height: 169.6 ± 8.8 cm; mass: 64.5 ± 16.9 kg). Participants were in good general physical health with no history of lower back or hip injury/pain within the past 12 months. Participants provided written informed consent prior to study initiation. Participant stature range was matched across percentage of stature for the general population between genders (15^{th} percentile to 90^{th} percentile stature for both genders). This stature range matching was an attempt to eliminate a stature effect and to represent even portions of the full scale of the general population.

5.2.2 Experimental Setup

Participants kneeled in a custom-built axial twist jig adapted from the frictionless table used in Drake & Callaghan (2008). The participants' thorax/upper body and pelvis/lower body were each secured using separate adjustable frame and harness systems (Figure 5.1A). The upper body system consisted of a padded posterior plate and adjustable nylon shoulder straps that wrapped anteriorly from above the shoulder and under the axilla. Straps were tensioned as tight as was comfortable for each participant. The lower body system consisted of a rigid wooden frame fixed to a padded kneeling cradle, anterior pelvis support and posterior gluteal plate. The gluteal plate was tensioned as tight as comfortable for each participant using a pair of threaded rods through the pelvic plate. The upper body frame was secured to the wall and allowed for sagittal plane (flexion-extension) movement, but restricted twisting and lateral bending movements. The lower body cradle could freely translate and rotate (axial twist) on nylon ball bearings (Salem Specialty Ball Inc., Canton, Connecticut, USA) scattered on the surface of the frictionless table. The upper and lower restraint systems minimized contributions of the thorax, pelvis, and lower limbs to trunk axial rotation movements and isolated such movements to the lumbar spine.

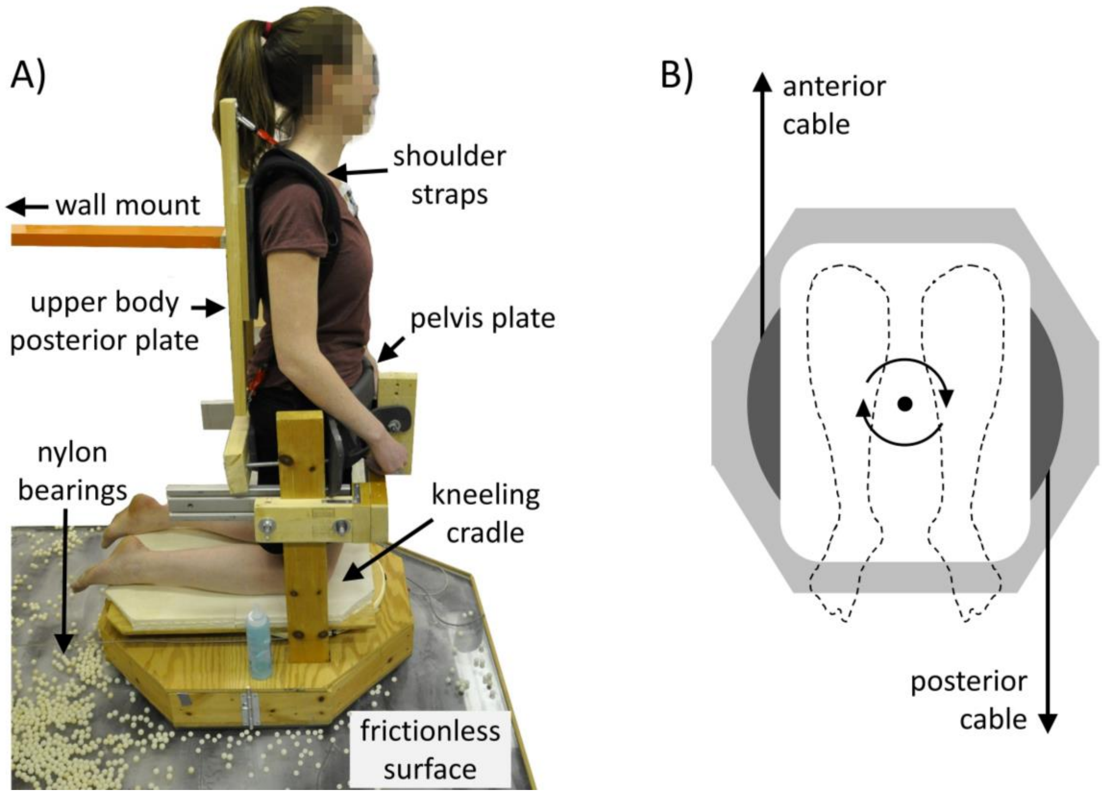


Figure 5.1: Custom-built jig used to create passive lumbar axial twist motion. A) The upper body and lower body harness systems and kneeling cradle, which freely rotates on nylon bearing on the surface of a frictionless table; B) cable system to create passive axial twist motion of the lower body system about the upper body system.

The lower body frame was rotated using cables attached to the anterior-left and posterior-right and wrapped around a cable guide (Figure 5.1B). A linear load cell (MLP-150, Transducer Technologies, Temecula, CA, USA) and signal conditioner (3270, Daytronic, Miamisburg, OH, USA) were attached in series with the cables to measure applied tension and calculate applied moment. Cables were pre-tensioned with the participant in the axial twist jig and pulled in parallel such that increased cable tension created axial twist motion of the lower body system about the upper body system.

5.2.3 Experimental Protocol

The skin surface above the spinous processes of the first lumbar to first sacral vertebrae (L1 to S1) was palpated and marked on the participant while lying prone prior to experimental trials. The lower body cradle height was adjusted to anterior superior iliac spine (ASIS) height, and the participant was inserted into the axial twist jig. Participants were instructed to remain passive with relaxed trunk musculature throughout all motion and measurement trials. Pilot data analysis indicated that participants were able to maintain lumbar extensor and oblique abdominal muscle activation levels below 5% of maximum voluntary contraction with this instruction.

Participants performed three repetitions of a maximum passive axial twist range of motion (ROM) trial as controlled by increased jig cable tension. The maximum thoracopelvic angle across the three trials was used as maximum ROM. For experimental trials, posture recordings were performed in each of four axial twist postures: 1) 0% ROM (neutral), 2) 25% ROM, 3) 50% ROM, and 4) 75% ROM. For each posture, cable tension was adjusted until real-time thoracopelvic angle reached the target angle. A static sonogram was recorded for each of six vertebral levels (L1 to L5 and S1), and thoracopelvic angle and cable tension were synchronously recorded. These six trials were recorded in each of the four axial twist postures for 24 total experimental trials. Applied axial twist moment about the pelvis was calculated from applied tension values and real-time body and jig kinematics.

5.2.4 Ultrasound Imaging

Posterior, transverse plane lumbar spine sonograms were recorded at six vertebral levels in each posture using a 13-6 MHz linear ultrasound transducer (8 cm footprint, 15 cm maximum depth) and a medical ultrasound system (M-Turbo, Sonosite, Bothel, WA). Each sonogram was scaled to dimensions in millimetres using a calibration image and a linear calibration procedure. Penetration depth, resolution, and intensity were kept constant for all sonograms within each participant.

5.2.5 Motion Capture

Participant and ultrasound transducer motion were recorded using an optoelectronic motion capture system (Optotrak Certus System, Northern Digital Inc., Waterloo, ON, Canada). Motion capture markers were attached to rigid plates which were affixed to the skin on the chest and to the lower body frame. A rigid body marker cluster was also attached to the dorsal surface of the ultrasound probe. Virtual markers were created by digitizing six locations on the thorax, six locations on the pelvis, and three points on the ultrasound transducer, based on a fixed rotational and translational relationship with the rigid body affixed to each of those segments. Virtual markers consisted of the sternal notch (SN), xyphoid process (XP), C7 spinous process (C7), T8 spinous process (T8), and bilaterally at the acromion (LAC, RAC), iliac crest (LIC, RIC), anterior superior iliac spine (LASIS, RASIS), and posterior superior iliac spine (LPSIS, RPSIS). Target data were recorded synchronously with sonogram data throughout all trials at 32 Hz. Pilot data analysis showed a mean thoracopelvic axial twist measurement difference of 1.19 ± 0.78 degrees when using a rigid body attached to the lower body jig instead of directly on the pelvis. This represents approximately 3% of maximum thoracopelvic axial twist observed in other studies (Drake & Callaghan, 2008; Gomez et al., 1991). Due to this low error and obstruction of pelvis markers from the lower body cradle, virtual markers relative to the lower body cradle rigid body were used for data collection and analyses.

5.2.6 Vertebral Angle Definition

Vertebral axial twist was characterized as the relative axial twist of each vertebra about the pelvis using a custom-written image analysis tool in MATLAB (2015a, Natick, MA, USA). For each vertebral level, three boney landmark locations (generally the tip of the spinous process and two lateral laminar surface points) were digitized on the neutral (0% ROM) sonogram to create the vertices of a landmark triad. This triad was then mapped onto the twisted posture sonograms for the same vertebral level and rotated and translated to align with common boney landmarks (Figure 5.2). Boney landmark vertex locations positions were digitized three times by the same experimenter in order to calculate intra-rater error.

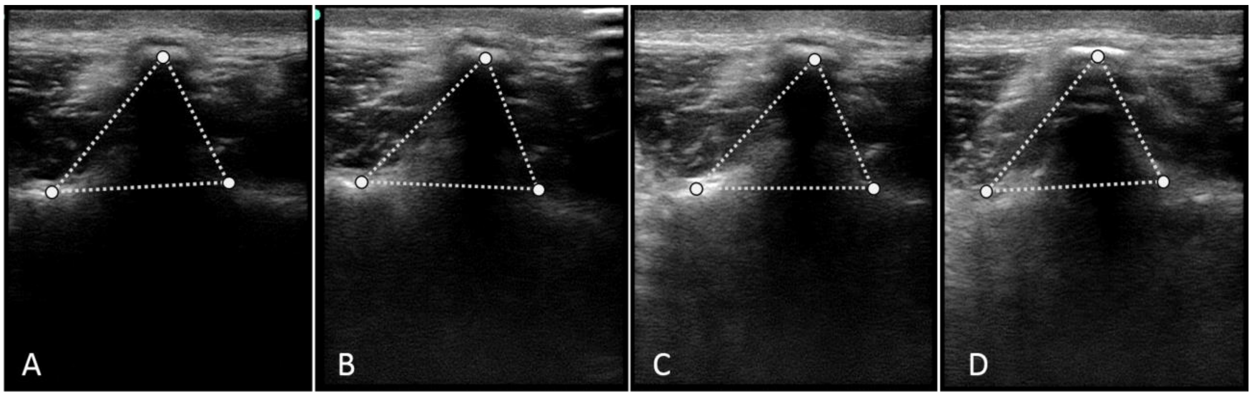


Figure 5.2: Digitized vertices on neutral posture sonogram (A) and mapped onto (B) 25% ROM, (C) 50% ROM, (D) 75% ROM twisted postures. Sonogram points were consistent boney landmark locations. Landmark triads were rotated and translated to align with common landmarks on each sonogram.

5.2.7 Local Coordinate System Definition

Local coordinate systems (LCS) were developed for the thorax, pelvis and ultrasound transducer. Thorax and pelvis systems used International Society of Biomechanics (ISB) conventions for spine motion (Wu et al., 2002) to define local system axes. The ultrasound transducer LCS (z-axis: along transducer face; y-axis: along handle length; x-axis: perpendicular to y- and z-axis) defined the sonogram location in global space, and allowed transformation of triad positions into the global system axes.

5.2.8 Data Reduction and Statistical Analysis

Three-axis Euler angles between the thorax and pelvis segments and axial twist of each vertebral triad about the vertical pelvis axis were calculated for each vertebral level in each posture. All thoracopelvic angles were normalized to a static, neutral trial and triad axial twist is expressed relative to the neutral (0% ROM) trials.

To evaluate the effect of vertebral level on vertebral axial twist (VAT) across thoracopelvic postures, a mixed general linear model (vertebral level (LEVEL); measured thoracopelvic angle (TP angle); sex) with participant as a random effect was used with statistical significance set at α = 0.05. To evaluate the relationship between the external TP and interval VAT angles, a linear regression was performed for each vertebral level with y-intercept forced through zero, and R-squared and mean error from the linear fit were calculated. All statistical analyses were performed using R-Studio (v1.0.136, Boston, MA, USA). All results are presented as mean ± standard error, unless otherwise indicated.

5.3 Results

Maximum thoracopelvic axial twist range of motion was 41.1 ± 1.8 degrees (range: 33.1 to 57.0 degrees). Mean flexion-extension angle relative to neutral was 1.94 ± 0.18 degrees of extension across all trials, with a maximum of 3.12 ± 0.35 degrees of extension at 50% of maximum axial twist range of motion. Mean lateral bending angle relative to neutral was 0.21 ± 2.4 degrees (leftward) across all trials, with a maximum of 0.6 ± 2.9 degrees (leftward) at 50% of maximum axial twist range of motion. A significant *TP angle*-by-*LEVEL* interaction effect (p<.0001) and a significant TP angle main effect (p<.0001) were observed for VAT angle. Angle increased superiorly (Figure 5.3) with L2 showing greater axial twist than L3 (p=0.009) and L5 showing greater axial twist than S1 (p=0.048). All other adjacent intervertebral comparisons (L1-L2, L3-L4 and L4-L5) were not statistically different across the measured TP angle range.

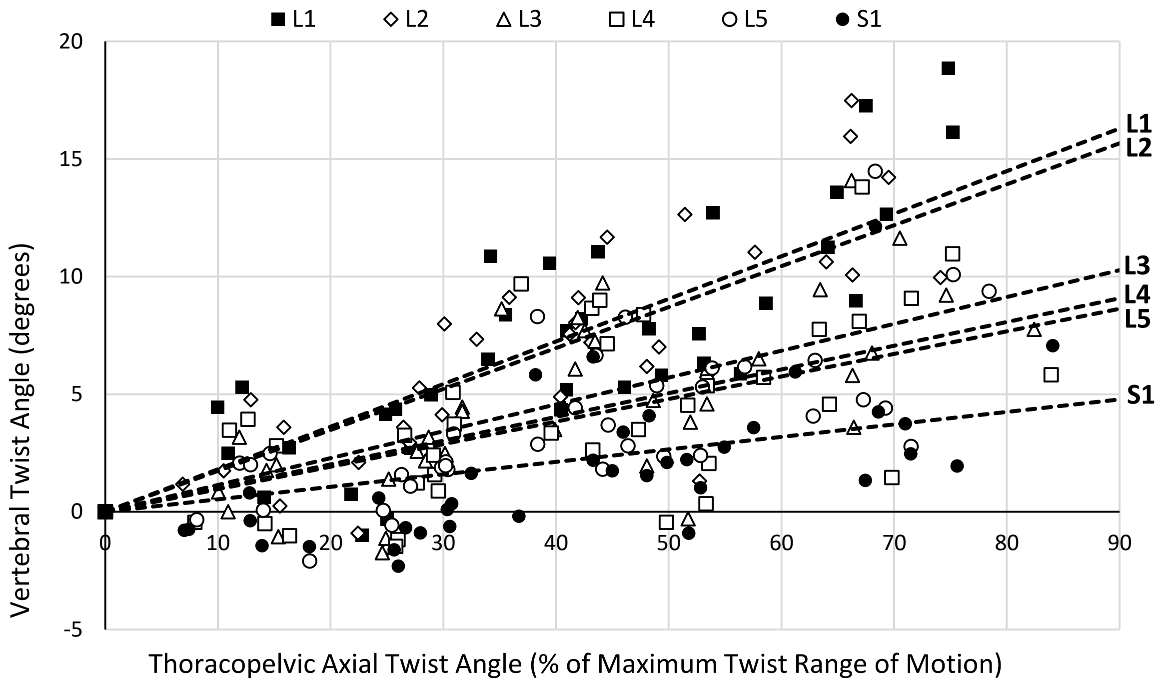


Figure 5.3: Thoracopelvic by vertebral axial twist angle for each vertebral level from L1 to S1. Linear regression lines are plotted and marked for each vertebral level.

Linear regression R-squared across vertebral levels increased superiorly from S1 to L1 and ranged from 0.427 to 0.785. Mean error from the linear regression fit varied across levels and ranged from 0.24 ± 0.3 degrees at L3 to 0.61 ± 0.5 degrees at L2 (Table 5.1). The total lumbar axial twist range of motion across all measured levels was 12.8 degrees based on the L1 linear regression fit at 100% of thoracopelvic range of motion. This lumbar motion accounted for 31.1% of total thoracopelvic range of motion. Mean intra-rater error across three iterations of boney landmark digitization was 0.62 ± 0.36 mm in the anterior-posterior direction and 0.56 ± 0.31 mm in the mediolateral direction.

Table 5.1: Linear regression R-squared, error and slope by vertebral level.

	Vertebral Level					
	L1	**L2**	**L3**	**L4**	**L5**	**S1**
R-squared	0.785	0.747	0.621	0.491	0.615	0.427
Error (degrees)	0.25 ± 0.3	0.61 ± 0.5	0.24 ± 0.3	0.29 ± 0.3	0.13 ± 0.3	0.24 ± 0.3
Slope	0.181	0.174	0.114	0.101	0.096	0.053

Error: Difference between actual data and linear fit line (mean ± standard error).

5.4 Discussion

The results demonstrate that segmental axial twist can be measured with moderate to high correlation in relation to external thoracopelvic twist using an ultrasound imaging technique to track boney landmarks in the lumbar spine. This study demonstrates how each vertebral level contributes to lumbar spine motion during thoracopelvic axial twist movements.

The majority of axial twist motion occurred between L2-L3 and L5-S1. These intervertebral joints accounted for 46.8% and 33.5% of the total lumbar spine motion, respectively. Fundamental multi-axis spine modelling studies have historically used L4-L5 as the primary site of motion and as a representative joint for the lumbar spine or low back (Marras & Granata, 1995; McGill, 1991). Studies examining intervertebral flexion angles have justified this by reporting much greater motion at the L4-L5 level (13 degrees) than the L1-L2 level (5 degrees) (De Carvalho et al., 2010). The current study indicates that L5-S1 would be a more accurate representation of lower lumbar spine motion in axial twist modelling, or perhaps that individual vertebral level modelling is required for high fidelity twist representation.

The overall fit of the individual vertebral axial twist measures to a linear regression increased superiorly. This best fit was at the most superior level (L1, R-squared: 0.79) and the worst was at the most inferior level (S1, R-squared: 0.43). These differences in fit by level are likely driven by individual variability. The vertebral motion across thoracopelvic range was generally not consistent for all participants at any of the measured vertebral levels. For Example, L1 twist was generally linear across the range of motion, but some participants had less than 15% of total L1 twist before 50 %ROM and the remaining 85% of motion after than point. This type of variability was consistent across vertebral levels, and indicates that level-by-level intervertebral axial twist motion exhibits a great deal of individual variability. Previous studies quantifying axial twist range of motion have found similar variability within their participant populations (Haughton et al., 2002; Ochia et al., 2006; Pearcy & Tibrewal, 1984a), and have presented differing distribution of motion across vertebral levels. Participant posture may play a role in these distribution differences, as some studies have been upright and others supine. An upright posture increases the compressive spine load, which decreases inter-facet spacing and increases resistance to axial twist (Drake et al., 2008), but ultimately reduces the strength of the spine (Aultman et al., 2004). These altered facet interactions play a significant role in intervertebral motion and range of motion (Drake & Callaghan, 2008).

The magnitude of thoracopelvic axial twist associated with the vertebral joint twist also contributed to both the individual variability and any differences between measured values from the current study and previous investigations. Thoracopelvic axial twist range of motion was 41.1 ± 1.8 degrees with a range of 33.1 to 57.0 degrees. This range was slightly greater than similar measures of 36.2 degrees and 39.4 degrees from a normative database (Gomez et al., 1991) and from a study involving a restraint system similar to the current study (Drake & Callaghan, 2008), respectively. Of the most recent investigations using medical imaging equipment, one did not report the applied thoracopelvic axial twist (Haughton et al., 2002), and the other reported only the amount of twist in the participant fixation apparatus (Ochia et al., 2006). The amount of thoracopelvic motion is critical for interpretation of vertebral level motion, and absence of this measure makes comparison to these previous investigations difficult. Future investigations attempting to evaluate vertebral axial twist motion should record and report associated thoracopelvic motion.

The lumbar spine is clearly not the sole contributor to total thoracopelvic motion. The total lumbar axial twist range of motion in the current study was an estimated 12.8 degrees using the linear regression for L1 and extrapolating to 100% ROM, which closely agrees with the 12.0 degrees maximum axial twist range of motion in cadaveric human spines (Pearcy & Tibrewal, 1984a). This lumbar motion accounts for just over 30% of the total thoracopelvic range of motion. Several studies have reported similar findings with aggregate lumbar angle accounting for around 20% of externally measured rotation range of motion in MRI and CT images of human participants (Haughton et al., 2002; Ochia et al., 2006). Further, the thoracic spine has been suggested to produce more movement in rotation than the lumbar spine (Fujii et al., 2007), and can account for over 40 degrees of axial twist during maximal thoracopelvic rotation tasks (Willems et al., 1996).

The flexion-extension posture may have also contributed to differences between the current study and previous evaluations of vertebral axial twist motion. The current study demonstrated that participants were able to hold a neutral sagittal posture with a mean thoracopelvic flexion-extension angle of 1.9 degrees (extension) from a neutral, upright posture. Other studies have observed over 5 degrees of flexion-extension difference across the lumbar spine or generally have not reported flexion-extension motion. Coupled axial twist and flexion-extension postures have been demonstrated to decrease stiffness (by 81%) and increase twist range of motion (14%) during maximal flexion, and to increase stiffness (125%) and decrease twist range of motion (24%) during maximal extension (Drake & Callaghan, 2008). These coupled posture differences present further challenges for predicting vertebral axial twist from externally measured postures and motion.

This measurement technique for vertebral axial twist demonstrated accurate measurement of vertebral axial twist in relation to thoracopelvic axial twist. This technique has potential for clinical and research use to assess impaired motion of an individual intervertebral joint, and to assess how vertebral motion changes across thoracopelvic range of motion. While this study was able to develop linear regression equations with reasonable fits for each vertebral level, the potential for use as a posture prediction tool is limited to the setup parameters used in this study. While an increased sample size would likely improve the regression fits, individual prediction accuracy may not be sufficient for evaluation of joint level injury mechanisms. The current study suggests that vertebral motion differences associated with coupled motion and individual variability – even among a healthy, asymptomatic population – may limit the reliability of these regression fits in some applied, occupationally relevant postures or motions but highlights the ability to measure vertebral axial rotation during activities.

5.5 Conclusion

This study investigated *in vivo* motion in healthy individuals, and demonstrated that vertebral axial twist motion can be measured using an ultrasound imaging technique. This vertebral motion can be directly associated with externally measured thoracopelvic motion using a motion capture system and setup typical for biomechanics studies. The majority of lumbar motion occurred at the L2-L3 and L5-S1 vertebral levels, suggesting that classic use of L4-L5 to represent lumbar spine motion may not be accurate for axial twist.

CHAPTER 6

STUDY 3: THE EFFECT OF AXIAL TWIST ANGLE ON *IN VITRO* CUMULATIVE INJURY LOAD TOLERANCE: A MAGNITUDE-WEIGHTING APPROACH FOR AXIAL TWIST EXPOSURES

6.1 Introduction

Cumulative biomechanical variables, specifically axial twist (Kelsey et al., 1984; Marras et al., 1993; Punnett et al., 1991) and cumulative compression and shear loading (Kumar, 1990; Norman et al., 1998; Seidler et al., 2001, 2003), have been identified and evaluated as important risk factors for the development of low back pain and *in vivo* tissue damage. Several factors can influence cumulative load exposure, including total exposure time, loading magnitude, loading rate, repetition rate, and posture. Each of these factors needs to be understood to be able to predict cumulative loading during a given task or scenario, and to develop strategies to reduce cumulative loading and associated development of low back pain.

Previous work has evaluated the effect of load magnitude on cycles to failure and has indicated that vertebral compressive fatigue life decreases non-linearly with increasing magnitude of repetitive, compressive loads (Brinckmann et al., 1988; Hansson et al., 1987; Parkinson & Callaghan, 2007a). This relationship and a strong link between cumulative load and *in vivo* tissue damage (Seidler et al., 2001, 2003) elicited methods of cumulative risk calculations such that higher load magnitudes generated higher injury risk scores. Seidler *et al*. (2001) used a squared compression load and Jäger *et al*. (2000) used a fourth-order weighting of compression load, but did not identify a clear physiological basis for these factors. Parkinson & Callaghan (2007b) used a cumulative *in vitro* loading protocol with cyclic compressive loads at 40, 50, 70 and 90% of estimated compressive strength to evaluate failure tolerance changes. Load magnitude and cumulative load results were used to develop a tissue-based equation that derived physiologically-based weighting factors (WFs) for cumulative injury risk. Similarly, Howarth & Callaghan (2013) evaluated cyclic shear loads at 20, 40, 60 and 80% of ultimate shear failure force and derived two weighting-factor functions for the relationship between submaximal shear force and sustained cumulative shear force to failure. Such approaches may improve the ability to detect cumulative load risk in both laboratory and occupational settings.

Compressive and shear cumulative loading modifiers present unique load tolerance and distribution characteristics, and further research is required related to multi-axis lumbar spine loading. During a cyclic flexion-extension loading protocol, the addition of axial moment accelerated the susceptibility of the intervertebral joint to injury (Drake et al., 2005) and may alter the failure mechanism (Drake & Callaghan, 2008). The addition of 5 N·m of static axial moment elicited higher incidence of facet joint fracture, and higher energy dissipation (Drake et al., 2005). In contrast to compressive and shear loading, the effect of submaximal axial twist loading of specimen fatigue life has not yet been investigated, and weighting factors for cumulative injury risk during cyclic axial twist loading do not exist.

Twisting exposures have been repeatedly identified as a risk factor for occupational low back pain and injury. Given the vastly different failure mechanisms between cyclic compression (vertebral endplate fracture, Parkinson & Callaghan (2007b)), shear (pars interarticularis fracture, Howarth & Callaghan (2013)), and twist loading (facet fracture, Drake *et al*. (2005)), there is a need for an improved understanding of the role of axial twist magnitude and associated moment as modifiers of the cumulative load tolerance of intervertebral joints.

The purpose of this study was to mathematically characterize the relationship between axial twist motion and moment magnitudes and the cumulative load tolerance of porcine cervical functional spinal units. From this relationship, non-linear equations for deriving appropriate weighting-factors for estimates of cumulative axial twist exposure were developed.

6.2 Methods

6.2.1 Specimen Preparation

A total of 24 FSUs (12 C3-C4 and 12 C5-C6) were excised from 12 fresh-frozen porcine cervical spines (C1-T1). A porcine cervical model was used due to the anatomical and functional similarity to the human lumbar spine (Oxland et al., 1991; Yingling et al., 1999). An animal model also provides control of specimen age, weight, diet and activity level (Parkinson & Callaghan, 2007b). Specimens were thawed overnight and surrounding musculature and fat were removed, leaving the osteoligamentous spine comprised of two vertebrae, the intervertebral disc (IVD), and surrounding ligaments. Transverse anterior processes were also removed to mimic facet axial twist resistance in human spines. Following dissection, IVDs were assessed for degeneration and graded on the scale outlined by Galante (1967). All specimens used in this study were classified as grade 1 disc quality.

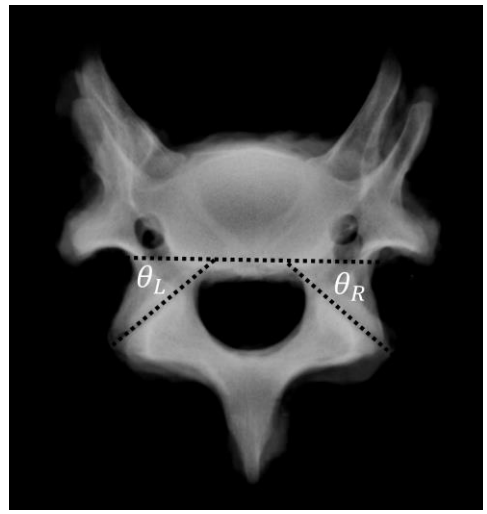


Figure 6.1: Transverse plane x-ray to determine left and right facet angles relative to the posterior aspect of the vertebral body.

Endplate area was calculated using the anterior-posterior depth (D) and mediolateral width (W) as per Callaghan & McGill (1995). The average area of the two exposed endplates was used as the endplate area of the FSU (Howarth & Callaghan, 2013; Parkinson & Callaghan, 2007b). As per Howarth & Callaghan (2013), bilateral facet angles were quantified using a transverse plane X-ray to determine facet tropism (Figure 6.1). Angles were determined relative to a line parallel to the posterior aspect of the vertebral body. Facet tropism was calculated as the difference between the left and right facet angles, and the degree of tropism was classified according to the scale described by Boden *et al*. (1996).

Specimens were mounted in custom aluminum cups using four screws (into anterior aspect of body; into lateral aspect of body; into exposed end plate) into the vertebral body (Figure 6.2A) and nonexothermic dental plaster (Denstone, Miles, Southbend, IN, USA). Specimens were aligned such that the midplane of the intervertebral disc was parallel to the surface of the cups. Specimens were wrapped in saline-soaked gauze to prevent hydration loss over the duration of the experimental protocol (Callaghan & McGill, 2001a). Threaded rigid body fixation screws were inserted into the anterior side of the superior and inferior vertebral bodies and extended vertically (Figure 6.2B). A rigid body containing four infrared markers was attached to the terminal end of each fixation screw and virtual marker positions on the FSU were recorded during the experimental protocol using an optical motion capture system (Optotrak Certus, Northern Digital Inc., Waterloo, ON) at 32 Hz.

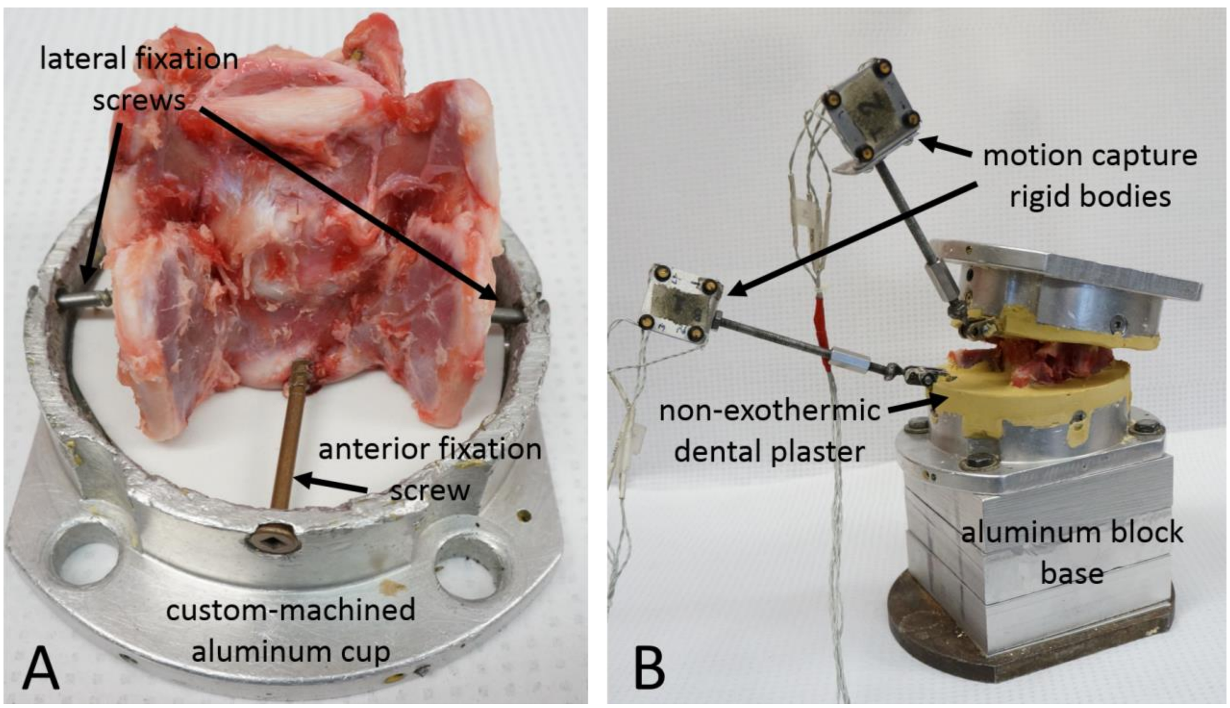


Figure 6.2: Fixation of a porcine cervical spine FSU in an aluminum cup testing assembly. A) Four screws (3 visible) were used to anchor the specimen to the aluminum cups, and B) dental plaster was used for further fixation and rigid bodies were attached to rods in the anterior vertebral body for motion tracking.

The specimen was mounted to an aluminum block base and inserted into a modified servo-hydraulic dynamic testing system (Model 8872, Instron Canada, Burlington, ON, Canada) that allowed load-control in the compressive axis, and position-control in the flexion-extension (FE) and axial twist (AT) axes. FE position was controlled using a servo-controlled torque motor and custom-built specimen carriage, and

AT angle was controlled using a steel rod threaded through the aluminum block base and a pair of servo-controlled linear actuators (RSA24, Tolomatic Inc., Hamel, MN, USA) driven by brushless servomotors (AKM22E, Danaher Motion Inc., Radford, VA, USA; Figure 4.2). Actuators moved in opposite directions to create a force couple about the centre of the FSU.

6.2.2 Loading Protocol

Prior to AT testing, FSUs were preloaded with a 300 N compressive load in a servo-hydraulic dynamic testing system (Model 8872, Instron Canada, Burlington, ON, Canada) set to zero flexion/extension moment for 15 minutes to counter any post-mortem intervertebral disc swelling (Callaghan & McGill, 2001a). The angular position at the end of the preload was used as the zero position for further testing of each specimen (Drake et al., 2005). A compressive load was applied to the FSU specimen, which corresponded to 15% of the ultimate compressive failure strength tolerance as calculated from a linear regression calculation based on endplate area (Parkinson et al., 2005).

Specimen flexion-extension range of motion was quantified using five repeats of a flexion-extension range of motion test at a rate of 0.5 degrees per second (Drake, 2008). The point of initiation of resistance to motion during flexion (when the moment versus angular position curve deviates from the initial linear section) was defined as the static flexion angle. This measure mimics the neutral zone defined by Panjabi *et al.* (1989) and the linear region described by Adams *et al*. (1980). Specimens were tested in axial twist at the midpoint of the neutral zone range of motion.

Specimens were randomly assigned to one of six levels (5, 7.5, 10, 12.5, 15, and 17.5 degrees) of rightward AT from the zero position. Twist magnitudes fall within the range observed as unable to produce failure during pilot testing (4 degrees) and maximum range of motion for porcine cervical spine FSUs (20.6 degrees, Drake (2008)). Twist direction has been reported to have no effect on peak twist failure moment, stiffness, or failure energy (Drake, 2008), so only rightward rotations were evaluated.

Specimens were repetitively loaded to their target AT angle, in angular twist control, and returned to zero position at a rate of 1 Hz until failure occurred or 21 600 loading cycles were performed. This loading cycle maximum is equal to the maximum number of cycles used in similar investigations of compressive (Parkinson & Callaghan, 2007b) and shear loading (Howarth & Callaghan, 2013), and has been suggested to be a realistic duration of loading – equivalent to one working day – without presence of physiological repair processes (Parkinson & Callaghan, 2007a). Applied loads were recorded with linear load cells (MLP-150, Transducer Technologies, Temecula, CA, USA) and signal conditioners (3270, Daytronic, Miamisburg, OH, USA) attached in series with the linear actuators and sampled at 32 Hz.

6.2.3 Post-Failure Analysis

Failure and the cycle at failure were visually identified by a step change in the cumulative applied axial twist moment curve for each FSU (Figure 6.3). This curve was smoothed (dual-pass Butterworth digital filter, 0.5 Hz cut-off frequency) and differentiated, and any peak greater than 0.05 N·m/s in the first derivative curve was identified as a step change. Preceding and following loading, transverse and sagittal plane x-rays were taken of the specimen. Specimens were further dissected into individual vertebrae and failure was manually assessed by the investigator.

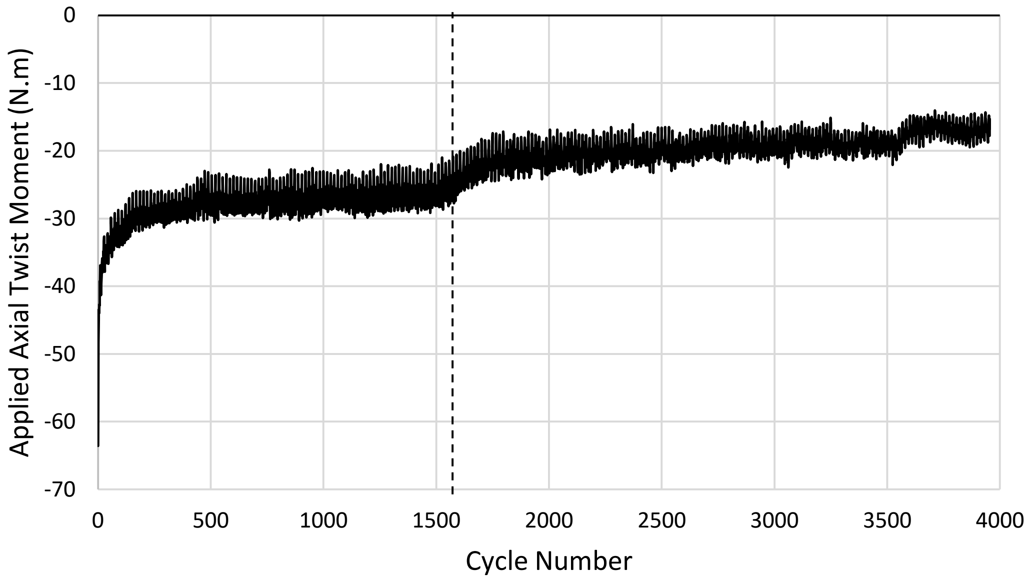


Figure 6.3: Example of step change in cycle-by-cycle net axial twist moment indicating the point of axial twist failure. Failure is identified by the vertical dashed line at cycle 1604.

6.2.4 Data Reduction and Statistical Analysis

Load cell voltages were linearly calibrated and converted to AT moment about the centre of the intervertebral disc. Moments were digitally filtered with a second-order dual-pass Butterworth filter using a filter cut-off frequency of 3.5 Hz (Howarth & Callaghan, 2013). Cumulative AT moment was calculated by trapezoidal integration of the moment versus time curve from the start of the loading protocol to failure or a maximum of 21 600 cycles.

6.2.5 Weighting-Factor Generation

Weighting-factors (WFs) were generated using a similar protocol described by Parkinson & Callaghan (2007b) and Howarth & Callaghan (2013). Two WF relationships were derived by relating AT angle magnitude to cumulative AT moment sustained prior to failure: 1) using the assigned actual AT angle magnitude, and 2) using the AT angle relative to maximum range of motion from Drake (2008). Identical data reduction, curve fitting, and threshold determination methods were used for generating both WF relationships.

Cumulative AT moment for only specimens that failed prior to the maximum cycle limit were used as data points for WF generation. An AT magnitude level was omitted from WF generation if none of the specimens assigned to that magnitude level failed prior 21,600 cycles. Continuous relationships between AT magnitude (absolute and relative to maximum range of motion) and cumulative AT moment sustained prior to failure were obtained by fitting a mathematical function to the datasets. Linear, power, and exponential functions were each fitted and the function with the greatest correlation (*r*) was used for data fit and weight-factor generation. The correlation value (0.89) for the exponential function (Equation 6.1) exceeded that of all other functions (0.67 to 0.84), and an exponential function was used for subsequent calculations.

$$C_{AT} = ae^{kx} \tag{6.1}$$

where C_{AT} was the calculated cumulative AT moment, a and k are coefficients derived by the exponential function fit, and x is the AT magnitude.

A WF of 1 was set as the minimum value and corresponded to the highest AT magnitude level where no specimens failed. This weighting factor value indicates that all AT magnitudes below this threshold (T) would not be adjusted when integrated to yield cumulative exposure in subsequent investigations. A value for T was determined for both WF relationships, and WFs were characterized using piecewise continuous functions (Equation 6.2)

$$w = \begin{cases} 1 & x \in R|[0, T] \\ \frac{1}{\alpha e^{kx}} + 1 & x \in R|[T, L] \end{cases} \tag{6.2}$$

AT magnitudes (x) below the defined threshold, T, were assigned a WF (w) of 1. AT magnitudes above the threshold were assigned a WF that was equivalent to the minimum WF (1) plus the inverse of the derived exponential function and coefficients. L is the maximum limit, with values of ∞ for absolute angle and 100 for magnitude relative to maximum range of motion.

6.3 Results

6.3.1 Post-failure Analysis

All specimens at the 5-degree AT magnitude, 2 specimens at the 7.5-degree magnitude, and 1 specimen at the 10-degree magnitude did not undergo visible failure. All remaining specimens underwent failure, with inferior right (compressive) facet fracture as the predominant injury in 16 of 17 (94.1%) specimens (Table 6.1). All specimens in the 17.5-degree group also exhibited a superior right facet

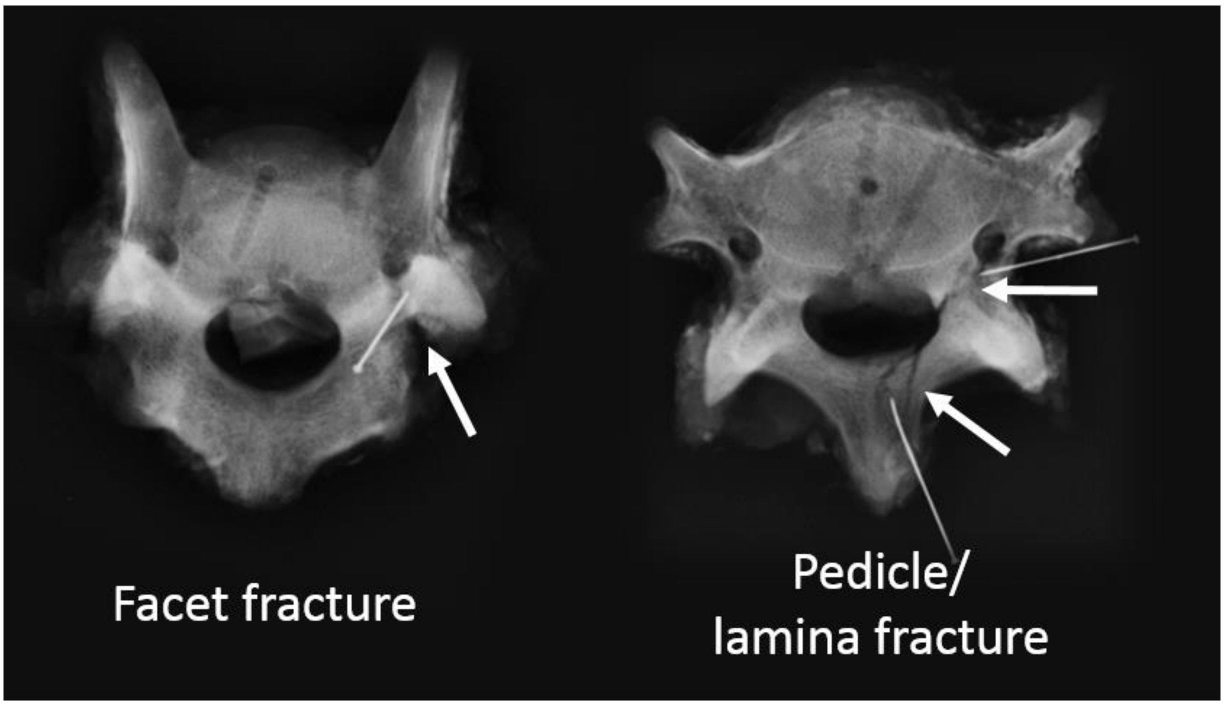


Figure 6.4: Illustration of predominant injury sites in specimens that failed during the 21 600 cycle protocol.

fracture in addition to the inferior facet fracture. All facet fractures occurred at the base of the facet (Figure 6.4), adjacent to the pars interarticularis. One specimen exhibited a combination lamina/pedicle fracture on the right side of the inferior vertebra. All specimens at the 17.5-degree and 15-degree AT magnitudes failed on the first cycle of the experimental protocol. Mean ±1 standard deviation facet angle was 46.5 ± 6.6 degrees and measured facet tropism was 1.1 ± 2.7 degrees with a range of 0 to 5 degrees across all specimens.

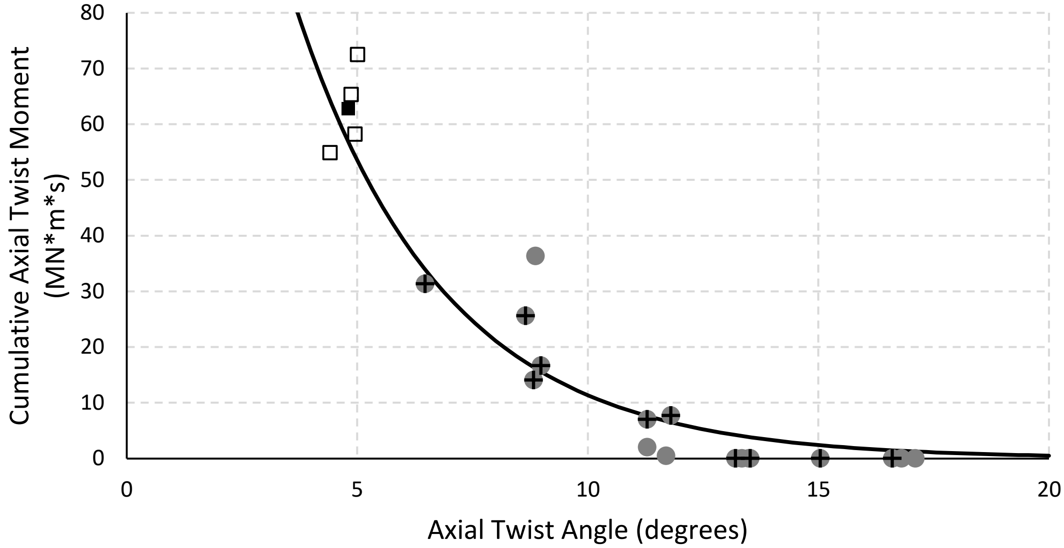
Table 6.1: Damage description for specimens that showed visible failure prior to 21 600 applied cycles.

Axial Twist (degrees)	Vertebral Level	Failure Description
17.5	C3-C4	Superior and inferior right facet fracture
17.5	C3-C4	Superior and inferior right facet fracture
17.5	C5-C6	Superior and inferior right facet fracture
17.5	C5-C6	Superior and inferior right facet fracture
15	C3-C4	Inferior right facet fracture
15	C3-C4	Inferior right facet fracture
15	C5-C6	Inferior right facet fracture
15	C5-C6	Inferior right facet fracture
12.5	C3-C4	Inferior right facet fracture
12.5	C3-C4	Inferior right facet fracture
12.5	C5-C6	Inferior right facet fracture
12.5	C5-C6	Superior and inferior right facet fracture
10	C3-C4	Inferior right laminar fracture
10	C5-C6	Inferior right facet fracture
10	C5-C6	Inferior right facet fracture
7.5	C3-C4	Inferior right facet fracture
7.5	C5-C6	Inferior right lamina and pedicle fracture

6.3.2 WF Generation

Since all specimens in the 5-degree AT group survived the 21 600 cycle protocol, data from those specimens were omitted from WF calculation. The minimum number of failed specimens in any remaining group was 2 at the 7.5-degree AT magnitude, so 2 data points were randomly selected from each remaining group to ensure equal contribution from each AT magnitude in curve fitting. This reduced dataset produced exponential functions between AT magnitude and cumulative axial twist moment that explained 89.4% of the variance for both the absolute AT angle (Figure 6.5A) and relative AT magnitude (Figure 6.5B) relationships. Sub-maximal thresholds, T, of 4.81 degrees and 22.7% ROM were used for WF calculations with Equation 6.2. Exponential function coefficients are expressed in Equation 6.3 and Equation 6.4 and illustrations of the piecewise continuous WF functions are presented in Figure 6.6.

A)



B)

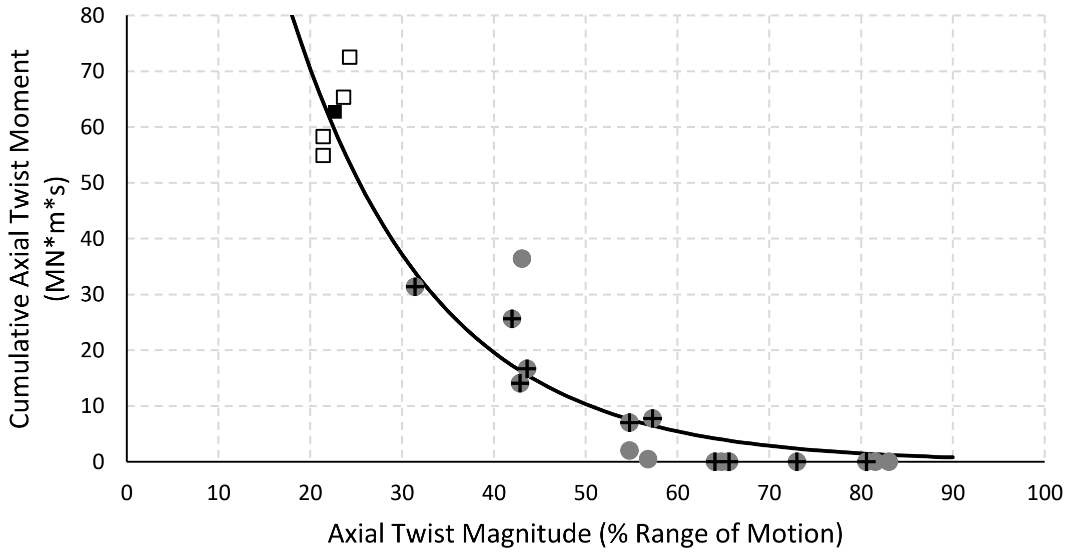


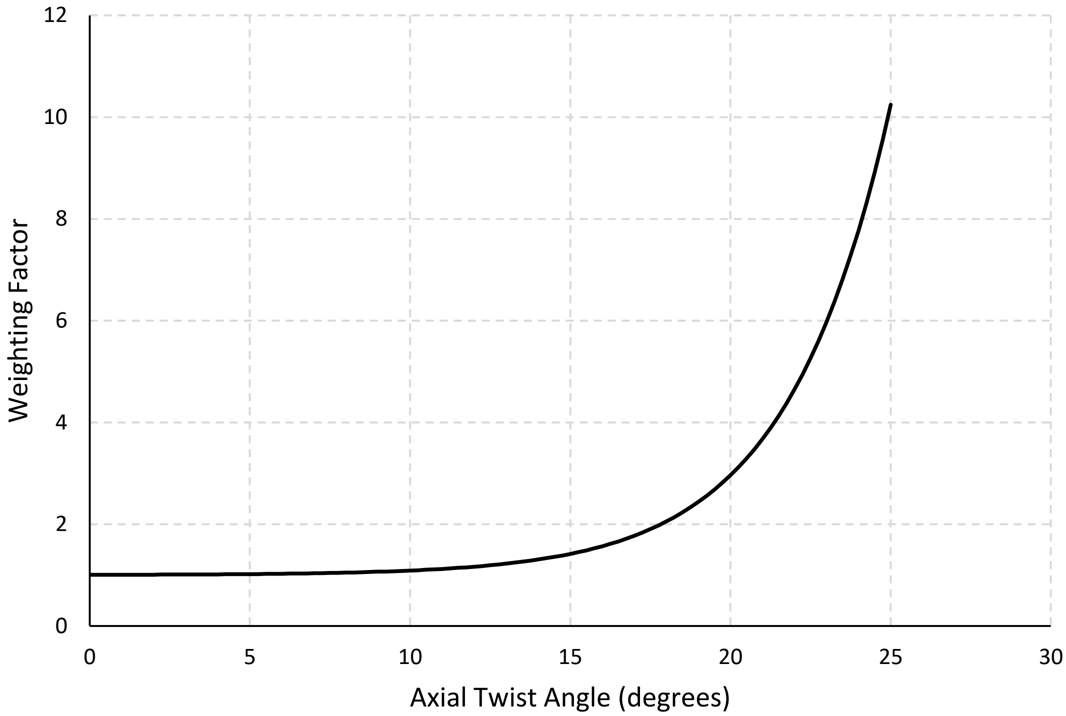
Figure 6.5: Exponential function (black lines) fits for calculating cumulative axial twist moment from axial twist magnitude in A) degrees, and B) percent of range of motion from Drake (2008). Gray circles indicate specimens that failed during the experimental protocol. Black crosses indicate specimens that were used for exponential fit calculations. Open squares indicate specimens that survived the entire experimental protocol, and the black squares are the means of all surviving specimens, which were used as the WF thresholds.

$$C_{AT} = 252.68e^{-0.064x} \quad \textit{absolute angle} \tag{6.3}$$

$$C_{AT} = 252.66e^{-0.310x} \quad \textit{relative to maximum range of motion} \tag{6.4}$$

where C_{AT} was the calculated cumulative AT moment and x is the AT magnitude.

A)



B)

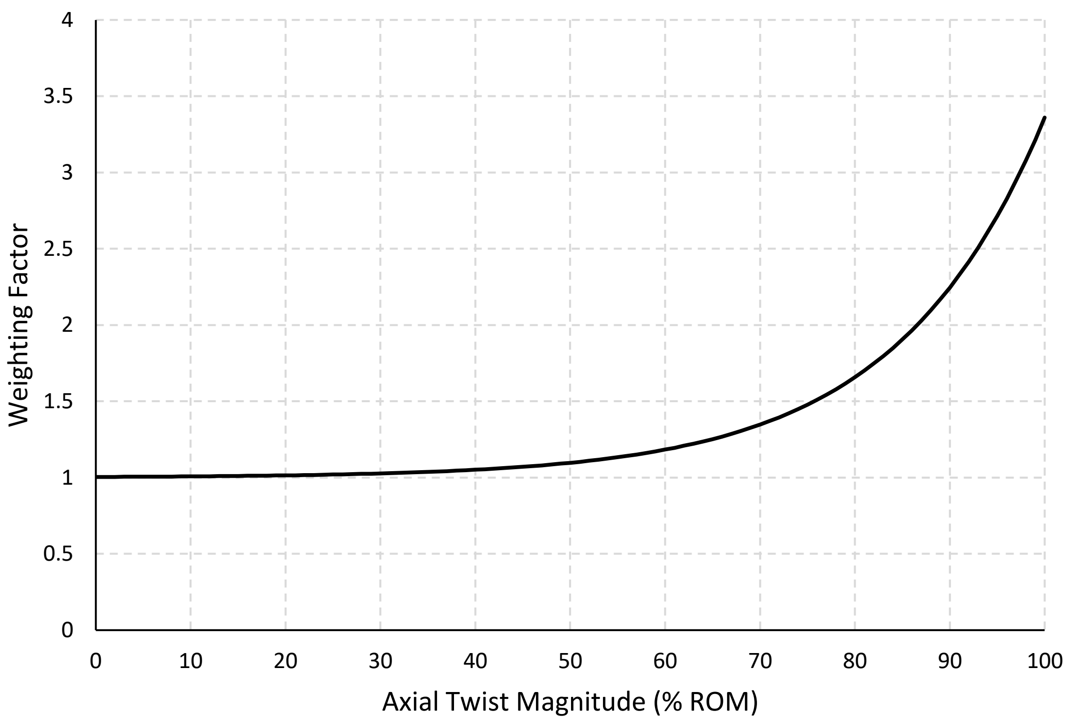


Figure 6.6: The WF functions for axial twist magnitude represented as A) absolute axial twist angle, and B) as a percentage of axial twist range of motion.

6.4 Discussion

Porcine cervical spines were used to mathematically characterize the relationship between axial twist motion and moment magnitudes and the cumulative load tolerance of porcine cervical functional spinal units in exponential functions. Weighting-factor functions for cumulative load were developed for neutral flexion-extension postures. This weighting-factor approach has the potential for use as an occupational tool for scaling cumulative low back injury risk related to AT based on magnitude of AT, either absolute or relative to range of motion.

The non-linear relationship between angle and moment indicated that a weighting-factor is required to properly assess risk related to repetitive or cumulative occupational twisting. This non-linearity of the weighting-factors and AT angle-cumulative AT moment relationships are common to similar studies of compressive (Parkinson & Callaghan, 2007b) and shear (Howarth & Callaghan, 2013) loading to failure. Prior to those studies, static cumulative load multipliers had been used (Hägg et al., 2000; Seidler et al., 2003), and indicated inaccurate estimates of the impact cumulative load has on failure tolerance. The non-linearity in the more recent studies illustrates the rapid change in injury risk with more extreme postures. Weighting-factors demonstrated a 10-fold increase across the absolute angular AT range of motion tested in this study and indicates that magnitude of AT has a direct influence on load tolerance in this axis of rotation.

Similar to the previously mentioned studies using non-linear weighting-factors, the current study was able to identify a threshold below which no failure occurred under the experimental conditions. This threshold (4.8 degrees and 22.7% of ROM) was a justified variable for the piecewise construction of the WF functions. Applied axial twist at or below these angular thresholds hold a WF value of 1 and should not have a significant effect on estimated cumulative axial twist moment tolerance. It should be noted that these thresholds apply to intervertebral motion, and do not necessarily linearly apply to aggregate lumbar spine or gross thoracopelvic motion. While methods for determining *in vivo* compressive failure tolerances have been developed (Davis & Parnianpour, 2003; Genaidy et al., 1993), similar relationships for lumbar spine axial twist remain underdeveloped. The use of an animal model in the current study is the first step towards assessment of injury risk weighting for cumulative axial twist moment.

Use of an animal model allowed precise control of axial twist magnitude over a long duration experimental protocol with low variability in angular position. This animal model provides an adequate testing specimen, as the frozen storage and thawing methods in this study have been shown to have very little effect on stiffness , hysteresis, or failure mode (Callaghan & McGill, 1995). In the current study, all specimens underwent identical thawing and preparation protocols, and any potential failure parameter changes are consistent across all specimens. The use of a porcine animal model allowed for control of age, size, and diet characteristics, and shares similar anatomy and function to a human lumbar spine (Oxland et al., 1991; Sheng et al., 2016; Yingling et al., 1999).

While a porcine cervical model has been used extensively to represent the human lumbar spine (Callaghan & McGill, 1995; Gardner-Morse & Stokes, 2003; Howarth & Callaghan, 2013; Parkinson & Callaghan, 2007b; Yingling et al., 1997), expressing the results from this study as a percentage of range of motion is essential for transferring results to human assessment. Thoracopelvic axial twist has been quantified in both seated and standing postures, and recent work had attempted to link this externally measured posture to internal, intervertebral geometry changes (McKinnon - Chapter 5; Haughton et al., 2002; Ochia et al., 2006). As these relationships develop, the results of the current study can be modified and applied to external thoracopelvic posture changes. However, maximum axial twist range of motion is similar between human lumbar (Farfan et al., 1970) and porcine cervical spine specimens (Drake, 2008), and absolute measures of axial twist in the current study may be directly applicable to human assessment. One possible limitation in this translation is the role of the anterior transverse processes of the porcine specimens. While they were removed for experimental testing, the role of the porcine spine anterior processed appears to be similar to that of the facet joints in both human and porcine models. It has been suggested that these processes may change the biomechanics of the FSU (Sheng et al., 2016) and could alter the resultant weighting-factors from this study compared to a human specimen. Future work is required comparing cumulative axial twist load tolerance in human and porcine specimens to confirm acceptable use of the weight-factors developed in the current study for human assessment.

Injury characteristics were very consistent across and within each of the evaluated AT magnitude groups. The predominant injury mode was fracture of the right side facet joint on the inferior vertebra. Given the inferior vertebra was twisted rightward relative to the superior vertebra, the failed facet was on the compression side. This failure mode is consistent with previous studies of axial twist to failure (Adams & Hutton, 1981; Drake, 2008), as it is suggested that the compression facet is the structure primarily responsible for the resistance of applied axial twist moment and motion (Adams & Hutton, 1981). The specimens in the 17.5-degree group additionally exhibited fracture of the right side facet on the superior vertebra. Facet geometry was not assessed during the experimental protocol, so it is difficult to speculate on the nature of this additional fracture site. Further, all fractures occurred on the first testing cycle, so it is unclear whether these fractures occurred simultaneously or sequentially.

The frequency of applied axial twist was held constant at a rate of 1 Hz for all specimens regardless of axial twist magnitude. This constant frequency affected the cycle-by-cycle load rate for each applied AT magnitude level, with a load rate 3.5 times higher in the 17.5-degree condition than in the 5-degree condition. While increasing load rate up to 10 degrees/second has presented no effects on failure or damage characteristics during axial twist, load rate has elicited varying responses in compression and shear loading. It is possible that the highest load rates in the current study (35 degrees/second) may have influenced facet recovery time and ultimate life to failure, but it has been suggested that the increased load rate with increased angular magnitude accurately reflects workplace exposures to match a fixed task cycle time. The 1 Hz frequency and 21 600 maximum number of cycles have been justified as a feasible number of loading cycles that would occur in 2 weeks of an industrial job at four cycles per minute

(Callaghan & McGill, 2001a; Drake et al., 2005; Howarth & Callaghan, 2013). These task parameters may be reasonable for accurate weighting of cumulative occupational injury risk, but future work should also investigate the role of higher loading rates on failure outcomes during axial twist to ensure axial twist angle magnitude is the predominant factor for risk weighting.

6.5 Conclusion

The results of this study demonstrated a non-linear relationship between axial twist angle magnitude and cumulative axial twist moment, and justify the use of a weighting approach to adjust injury risk related to cumulative load. The two mathematical relationships developed allow for application to a specific intervertebral joint angle in a clinical population, as well as the potential for use in assessment of occupational task postures normalized to maximum range of motion. Future work establishing the relationship between externally measured thoracopelvic posture and segmental intervertebral angle needs to be conducted to use these weighting-factor functions in human injury risk assessment. Once estimates of this relationship are established, cumulative axial twist moment exposure limits can more accurately improve occupational workstation and task design and mitigate twist-induced injury risk.

CHAPTER 7

STUDY 4: THE INFLUENCE OF HAND LOCATION AND EXERTION PARAMETERS ON LUMBAR AXIAL TWIST POSTURE DURING SIMULATED INDUSTRIAL TASKS

7.1 Introduction

Modern industrial workplaces have increasingly less manual materials handling, more assistive devices and automation, and controlled strategies for workstation design and occupational injury prevention. As a whole, these workplace strategies have appeared to have had a positive impact, as occupational lost time injuries in Canada have decreased every year since 2001 (Figure 7.1) and had a historical low 0.95 injury rate in 2013 (WSIB, 2014). However, lost time injury trends may be partially attributed to increased claim denials, rather than entirely due to injury reduction. Despite the overall decrease in accepted claims, high impact claims (including low back injury claims) remain abundant and represent over 40% of benefit payments (Figure 7.2) and over 35% of all lost time claims (WSIB, 2014).

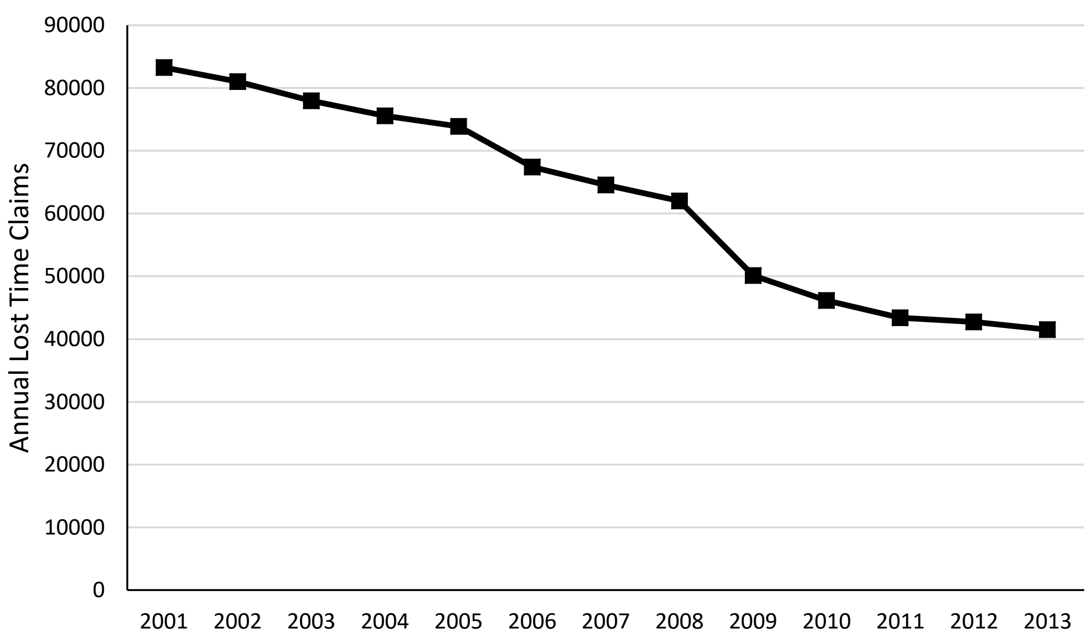


Figure 7.1: Annual registered lost time claims in Ontario by year.
Data from (WSIB, 2014).

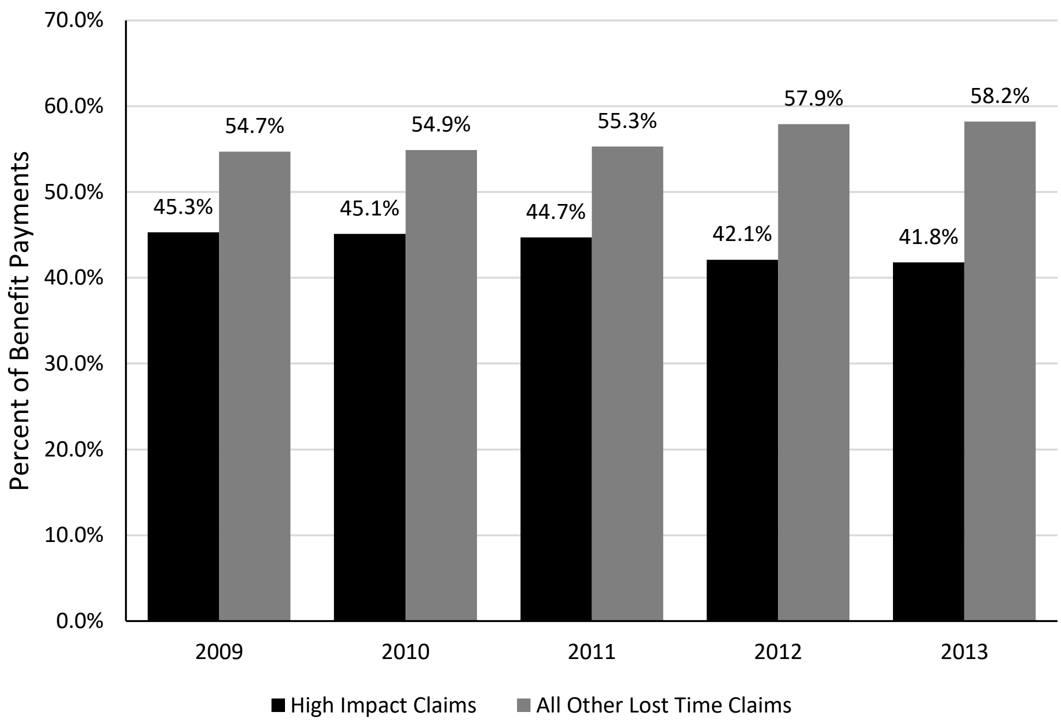


Figure 7.2: High impact lost time claims by benefit payment percentage over a five-year span (2009-2013), adapted from WSIB (2014).

Researchers have investigated factors that could lead to low back disorders, and have typically identified lifting heavy loads and awkward postures as primary risk factors. Since heavy loads are viewed as a straight-forward engineering control, early ergonomics related research targeted control of hand loads and developed limits for occupational low back loading (McGill et al., 1998; NIOSH, 1981). More recent review studies have identified that axial twist and twist combined with other motions are major potential contributors to low back disorder risk (Bernard, 1997; Hoogendoorn et al., 1999; Marras, 2000). It is believed that these types of motions and postures can be limited or eliminated through workstation and layout interventions, but such workstation design guidelines require research-based evidence of reduced associated injury risk. While guidelines related to low back tolerances (McGill et al., 1998; NIOSH, 1981), manual handling loads (Snook & Ciriello, 1991), and repetitive work thresholds (Kilbom, 1994; Norman et al., 1998) are abundant, postural guidelines are less common and are based on anthropometry and work dimensions (Ciriello, 2007; Keyserling et al., 1988) rather than cyclic joint loading and failure literature.

In practice, current workstation design guidelines include recommendations for forward and lateral reach distances during occupational tasks based on task frequency and anthropometry (CSA, 2012). Forward reach guidelines focus on maintaining a neutral shoulder posture, as flexed and forward reaching shoulder postures create large shoulder load moments (Anton et al., 2001; McClellan et al., 2009), show decreased time to reach fatigue (Brookham et al., 2010; Chaffin, 1973), and increased perceived exertion in workers (Dickerson et al., 2006). For lateral reaches, current guidelines are underdeveloped and do not have the same research-based foundation. Rather, lateral reach guidelines generally apply forward reach concepts to the lateral reach envelope determined by worker size and arm length.

The magnitude of axial twist in the lumbar spine in relation to reaching tasks is currently unknown. Therefore, the purpose of this study was to investigate lumbar spine axial twist during simulated occupational tasks across a range of forward and lateral reach distances, task heights, and exertion directions. The findings from this study will be used to assess low back injury risk related to the recommended reach distances for frequent, infrequent and occasional tasks. The physical outcomes from this study aim to enhance single-handed reach envelope recommendations in the context of spine motion and loading.

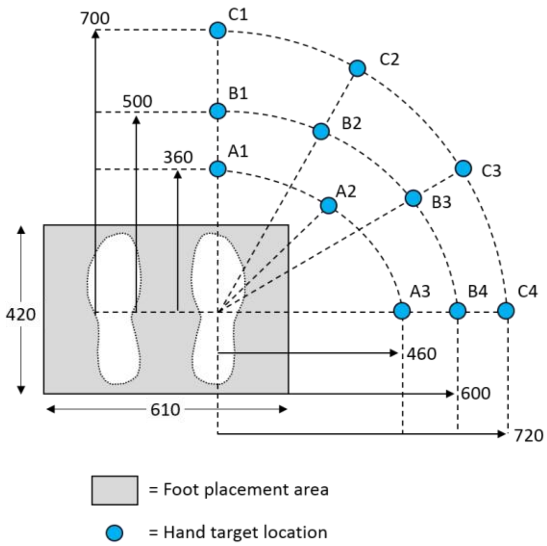
7.2 Methods

7.2.1 Participants

Twenty-four (12 male, 12 female) right-hand dominant participants aged 18-35 were recruited for this study. Hand dominance was determined by handwriting preference. Participants were in good general physical health with no history of lower back, hip, or upper limb injury/pain within the past 12 months. Participants provided written informed consent prior to study initiation.

7.2.2 Experimental Setup and Protocol

Participants performed single-handed exertions against an interface fixed to a load cell (MSA-6, AMTI, USA) which was attached to the end of a 6-degree of freedom robotic arm (Motoman HP50, Yaskawa, USA). Eleven right-hand target locations (Figure 7.3) corresponded to current ergonomic forward and lateral reach guidelines in primary, secondary, and tertiary reach zones for frequent, infrequent and occasional tasks, respectively (CSA, 2012). Lateral reaches of 460 mm, 600 mm and 720 mm and forward reaches of 360 mm, 500 mm and 700 mm corresponded with each of the three reach zones, respectively. Exertions were performed at each hand target location in all combinations of three exertion directions (forward push, upward exertion, downward exertion), two heights (standing acromion and olecranon heights), and two load magnitudes (40 N, 60% of forward push strength) for a total of 132 exertion trials. The 40 N hand load represented an occupationally relevant force level that would not produce fatigue effects (McDonald et al., 2012), and the strength-scaled load represented a psychophysically acceptable force (Fischer et al., 2012) in the limiting shoulder exertion direction (Chow & Dickerson, 2009). Trial order was fully randomized within each of the two task heights. For each trial, participants were asked to perform an exertion against the participant interface and maintain force at the target level for 3-seconds using a real-time force feedback display adjacent to the exertion interface (Figure 7.3). Feet were constrained within a 420-by-610 mm foot placement area (Figure 7.3) at shoulder width and with toes pointing anteriorly. Thoracopelvic and right upper limb posture and muscle activation data were recorded for 1-second during this static hold.



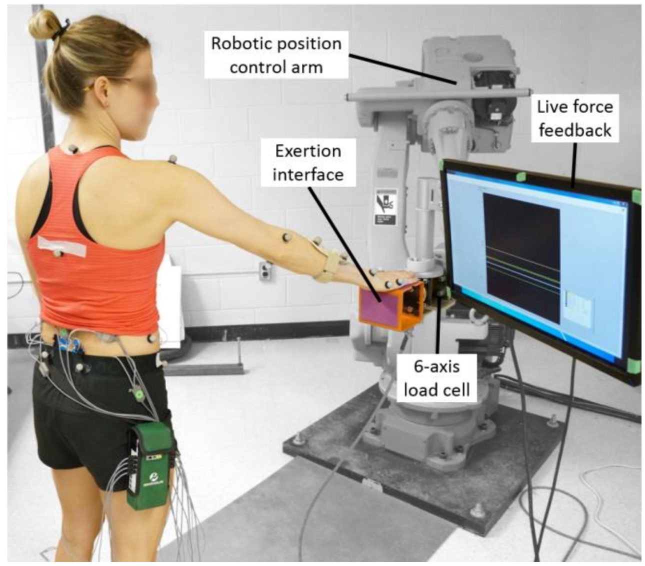


Figure 7.3: Illustration and photograph of experimental setup showing hand target locations relative to the right shoulder. *Left*: Hand targets were located at recommended distances for frequent, infrequent and occasional forward and lateral reaching tasks. All units are in mm. *Right*: participant exertion interface location was controlled by a robotic arm and force exerted on a load cell was displayed in real-time on a viewing screen.

7.2.4 Motion Capture

Participant thoracopelvic and right upper limb motion were recorded using an optical motion capture system (Vicon MX, Los Angeles, CA) and software (Vicon Nexus). Reflective motion capture markers were affixed to the skin at 18 boney landmarks on the thorax, pelvis, and right upper limb: suprasternal notch (SS), xyphoid process (XP), C7 spinous process (C7), T8 spinous process (T8), left and right acromion (LAC, RAC), left and right anterior and posterior superior iliac spines (LASIS, RASIS, LPSIS, RPSIS), left and right iliac crests (LIC, RIC), medial and lateral epicondyles of the humerus (ME, LE), radial styloid (RS), ulnar styloid (US), and the distal ends of the 2nd and 5th metacarpals (MC2, MC5). Additional segment tracking markers were placed on the right upper arm, right forearm, and right hand. Marker trajectory data were collected throughout all trials at 50 Hz.

7.2.5 Surface Electromyography

Participants were instrumented with surface EMG bilaterally over gluteus medius (GMD), internal obliques (IOB), lumbar erector spinae (LES; L3 vertebral level) and thoracic erector spinae (TES; T8 vertebral level). Electrode locations for the trunk muscles were determined as per McGill (1991), and the gluteus medius electrodes were placed slightly posterior to the proximal third of the line connecting the iliac crest and greater trochanter. Pre-gelled Ag–AgCl surface electrodes (Blue Sensor N, Ambu, Ølstykke, Denmark) were used with a 2 cm inter-electrode distance. Electrode placements were confirmed by palpation upon contraction. EMG signals were recorded using a telemetric system (Telemyo 2400T G2, Noraxon USA Inc., Scottsdale, AZ). System leads were equipped with a 1st order high-pass filter (10 Hz +/- 10% cut-off). Input channels had 10-500Hz analog band pass filters. EMG active lead specifications included a differential amplifier common mode rejection ratio of >100 dB and input impedance of >100 mΩ. Output signals were synchronously recorded with the motion capture data at 1500 Hz. Raw signals were linear enveloped using a single-pass, second-order Butterworth digital filter with a 2.5 Hz cutoff frequency (Winter, 2005). Smoothed signals were normalized to maximum voluntary isometric contractions (MVC). MVCs were obtained against manual resistance in the Beiring-Sorensen for both levels of erector spinae, using a modified sit-up for IOB, and using side-lying hip abduction for GMD (Nelson-Wong & Callaghan, 2010). Supine and rest quiet lying trials were recorded to verify resting activation.

7.2.6 Data Reduction and Statistical Analysis

Thorax, pelvis, and right upper arm segment local coordinate systems were constructed according to ISB conventions (Wu et al., 2002, 2005) – X: pointing forward, Y: pointing superiorly; Z: pointing right. Thoracopelvic angles (YZX Euler sequence) and right thoracohumeral angles (YXY Euler sequence) were calculated using custom-written Matlab software. Thoracopelvic axial twist and flexion-extension angles were normalized and expressed relative to a static upright standing trial. Pelvis axial twist angle was also calculated relative to a static, upright standing trial. Mean joint angles and muscle activations were compared using a mixed general linear model (RStudio 1.0.136) with *sex* (M/F), *reach* (A/B/C), *angle* (0 to 90 degrees), *height* (elbow/shoulder), *direction* (up, down, push), and *load* (absolute, scaled to strength) as factors (α = .05). These factor labels are used throughout the results section. A Tukey HSD post hoc test evaluated levels within significant main and interaction effects.

7.3 Results

Specific experimental results are presented for thoracopelvic (lumbar) axial twist, flexion-extension, and lateral bend posture, pelvis axial twist, thoracohumeral (shoulder) elevation, and trunk muscle activation. *Load* and *sex* showed no significant interaction or main effects for any measure (*p*>.388). As such, all results are collapsed across the two recorded load conditions and male and female participants. Results are presented as mean ± standard error.

7.3.1 Lumbar Posture

Axial Twist

Significant *reach*-by-*angle* (*p*=.003), *angle*-by-*direction* (*p*<.0001), *reach*-by-*direction* (*p*=.032) and *height*-by-*direction* (*p*=.028) interactions were observed for lumbar axial twist angle. Twist angle demonstrated a 7.3 ± 1.7 degree twist (17.8% ROM) to the contralateral side for straight forward hand targets (averaged across reach distances and exertions directions), and twist increased with angular distance from the midline of the body to a maximum of 14.7 ± 1.3 degrees (35.8% ROM) collapsed across exertion directions (Figure 7.4). This maximum reach was similar between reach distances (*range*: 14.4 to 15.0 degrees). Twist angle was greater for downward (*p*=.008) and push exertions (*p*<.0001) than upward exertions for hand target locations at 0-degrees (A1, B1, C1) and 30-degrees (B2, C2) from midline, and twist was greatest for push exertions for all other hand target locations (p<.0001,Figure 7.5). Twist was greater for exertions at shoulder height than at elbow height, with the greatest difference (8.5 vs. 4.5 degrees) for push exertions. No significant interaction or main effects for lumbar axial twist were observed involving sex or load.

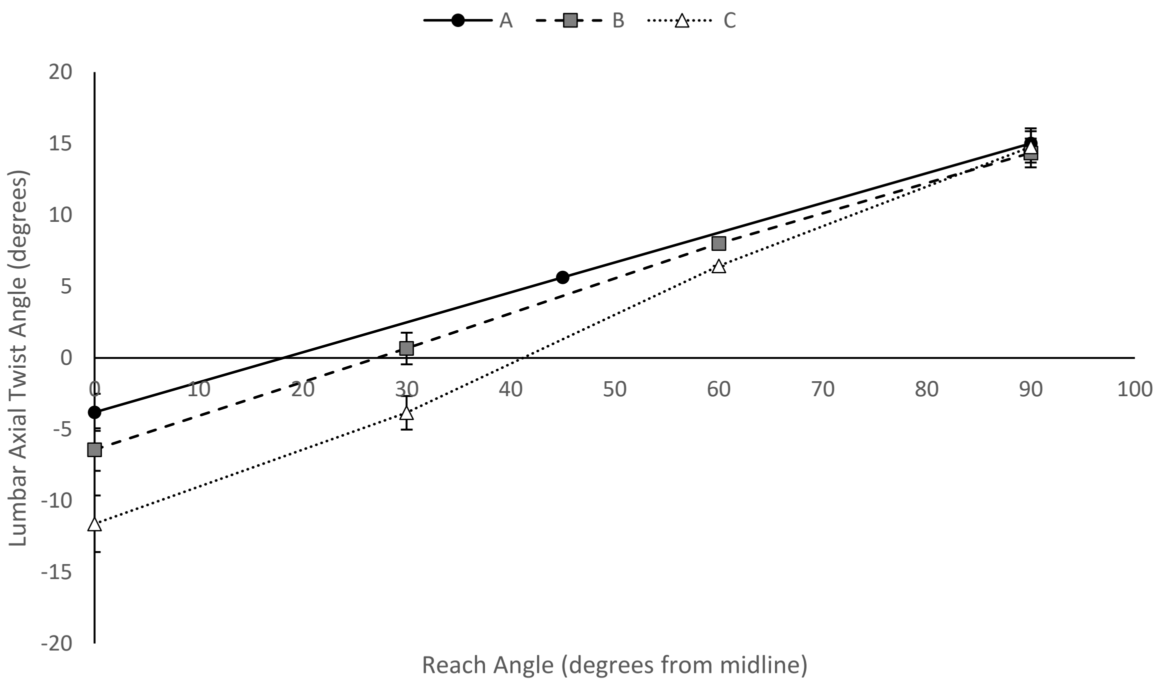


Figure 7.4: Lumbar axial twist angle by reach distance across hand target angle collapsed across sex, height, and load. Twist angle increased with increasing hand target. Positive values represent twist to the right. Error bars represent standard error.

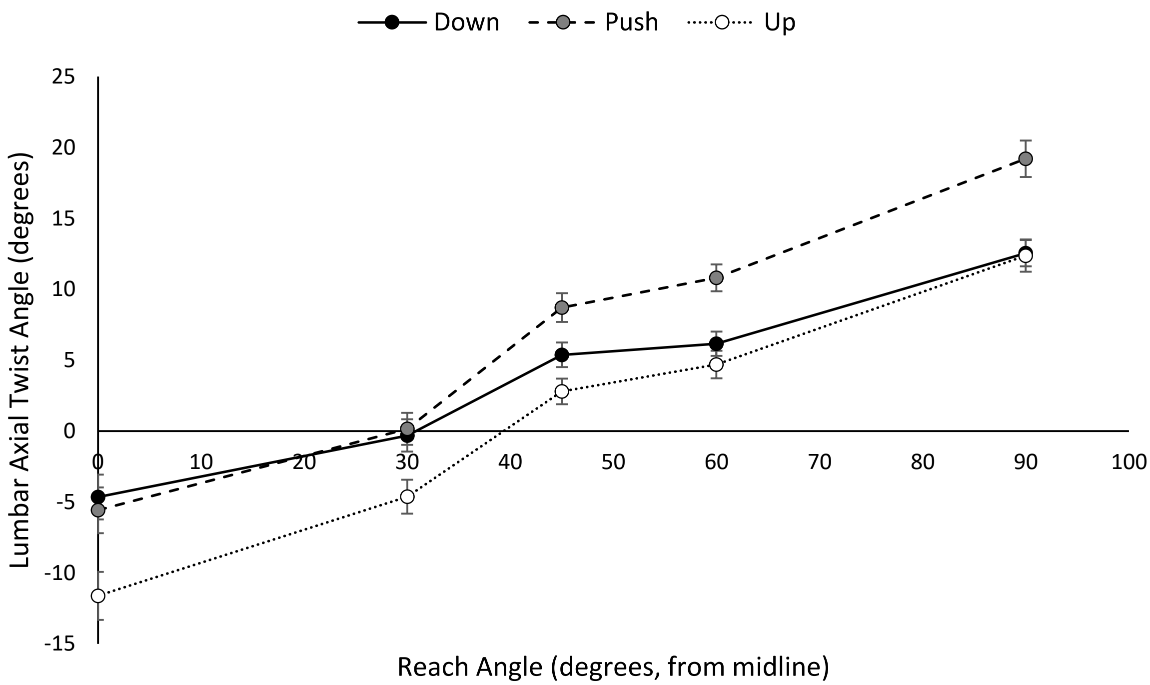


Figure 7.5: Lumbar axial twist angle by exertion direction across hand target angle collapsed across sex, height, and load. . Positive values represent twist to the right. Error bars represent standard error.

Flexion-Extension

A significant *direction*-by-*height* (p=0.008) interaction effect and *height* (p<.0001) and *direction* (p<.0001) main effects were observed for lumbar flexion-extension angle. Lumbar flexion was greater for push (p=.0001) and upward exertions (p=.0001) than downward exertions at elbow height. Flexion differences in exertion direction differed by hand target location with the greatest flexion during push and upward exertions at directly forward target locations. Flexion angle was 2.6 ± 1.3 degrees across all measures, and was greater at elbow height (5.0 ± 1.4 degrees) targets than shoulder height (0.3 ± 1.1 degrees) targets (p<.0001).

Lateral Bend

A significant *direction*-by-*height* (p=0.008) interaction effect and *angle* (p<.0001), reach (p<.0001), and *direction* (p<.0001) main effects were observed for lumbar lateral bend angle. Lateral bend (rightward) was greater for upward and forward push exertions at elbow height (9.6 ± 1.5 degrees; 5.8 ± 1.4 degrees) than at shoulder height (2.6 ± 1.1 degrees; 0.8 ± 1.0 degrees). Lateral bend angle was 3.2 ± 1.4 degrees across all measures, and demonstrated greater magnitude at hand targets 0 and 30 degrees from midline (4.6 ± 1.3 degrees) than at any more lateral angle (2.0 ± 1.3 degrees). Lateral bend was also greater for tertiary reach zone targets (4.7 ± 1.5 degrees) than primary or secondary reach zone targets (1.8 ± 1.3 degrees; 2.6 ± 1.3 degrees).

7.3.2 Pelvis Posture

Significant *direction* (p=.0002), *angle* (p<.0001), and *reach* (p<.0001) main effects were observed for pelvis axial twist angle. Pelvis twist angle was greater for downward and push exertions than upward exertions. Pelvic twist decreased with reach distance from primary (4.2 ± 1.9 degrees) to secondary (2.1 ± 2.0 degrees) to tertiary (0.8 ± 2.2 degrees) reach zones hand targets. Twist angle increased from a contralateral twist (-4.6 ± 1.7 degrees) at directly forward hand targets to a maximum of 9.7 ± 2.1 degrees at hand targets 90 degrees from midline (Figure 7.6).

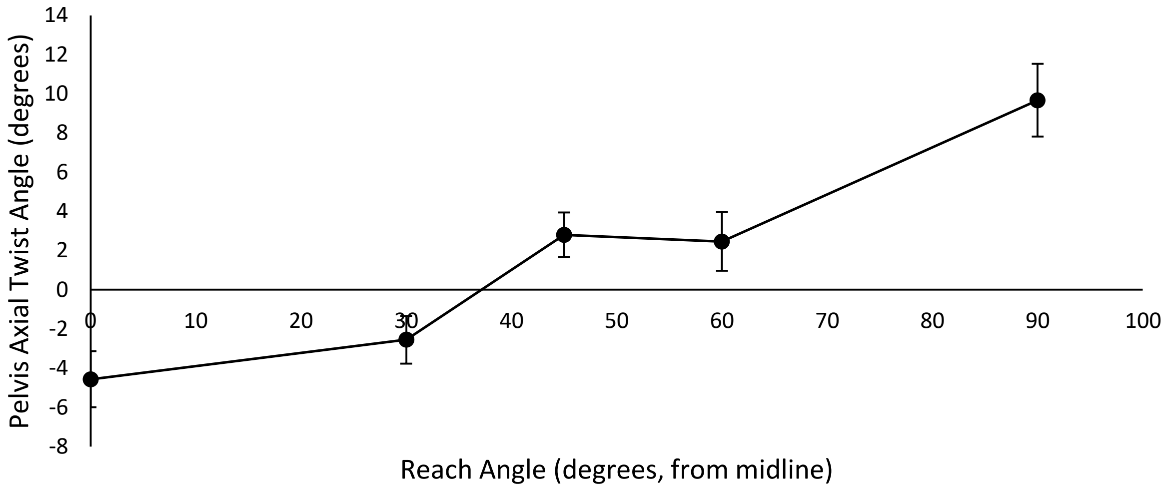


Figure 7.6: Pelvis axial twist angle by reach angle. Twist angle increased with increased laterality of hand location. Positive values represent twist to the right. Error bars represent standard error.

7.3.3 Shoulder Posture

Significant *direction*-by-*height* (p<.0001), *reach*-by-*height* (p<.0001), *angle*-by-*direction* (p=.026), and *angle*-by-*reach* (p=.021) interaction effects were observed for shoulder elevation angle. Elevation angle was generally highest for downward exertions (52.3 ± 3.8 degrees), and decreased slightly with increased laterality of hand target location (Figure 7.7). Elevation was greater for shoulder height tasks (50.0 ± 3.5 degrees) than elbow height tasks (28.3 ± 2.9 degrees) (p<.0001) and presented greater values for reaches in the tertiary (C) reach zone (48.9 ± 3.2 degrees) than the closer reach zones (p<.0001). Elevation decreased with increased laterality of hand target location within each reach zone. *Angle* (p<.0001) and *reach* (p<.0001) main effects were also observed with the same general trends.

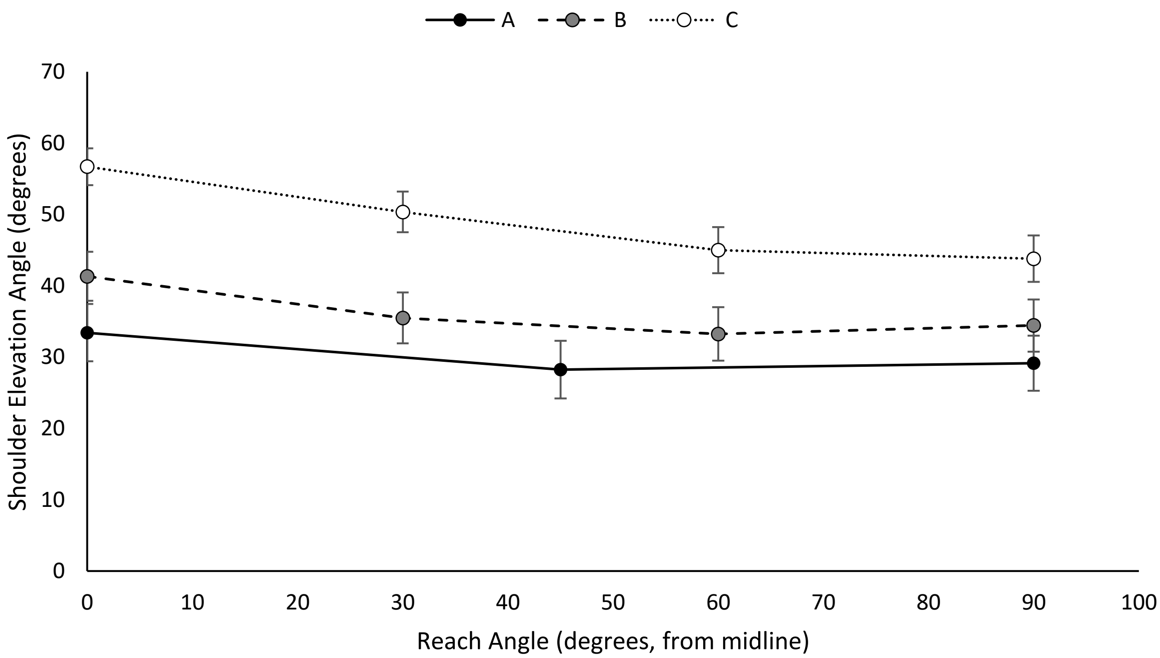


Figure 7.7: Shoulder elevation angle by exertion direction and hand target location collapsed across sex, height, and load. Elevation angle decreased with increased. Positive values represent elevation from downward vertical. Error bars represent standard error.

7.3.3 Trunk Muscle Activation

Left and right LES (p=.005, p<.0001) and left TES (p<.0001) each demonstrated an *angle*-by-*direction* interaction effect for mean activation. Lumbar level erectors exhibited increased activity in the tertiary reach zone and decreased with hand target laterality. Upward exertions (*left*: maximum 17.8 %MVC; *right*: maximum 15.4 %MVC) elicited greater activation than the other exertion directions (maximum 6.1 %MVC – push at C1) across all hand targets (p<.0001). Thoracic level erectors showed the same general trends except increased activation with laterality for forward push exertions (Figure 7.8). An *angle*-by-*direction* interaction effect for left IOB (p<.0001) and right GMD (p=.031) and an *angle* main effect for right IOB (p=.027) were also observed. Left IOB and right GMD activations were greatest for pushing exertions (15.0 ± 2.4 %MVC, 7.5 ± 1.4 %MVC), and activation increased for bilateral IOB and right GMD with increased laterality. Left GMD exhibited a *direction* main effect (p=.0004) with significantly greater activation in pushing (5.5 ± 0.7 %MVC) and upward (5.7 ± 0.6 %MVC) exertions than downward exertions (2.4 ± 0.3 %MVC).

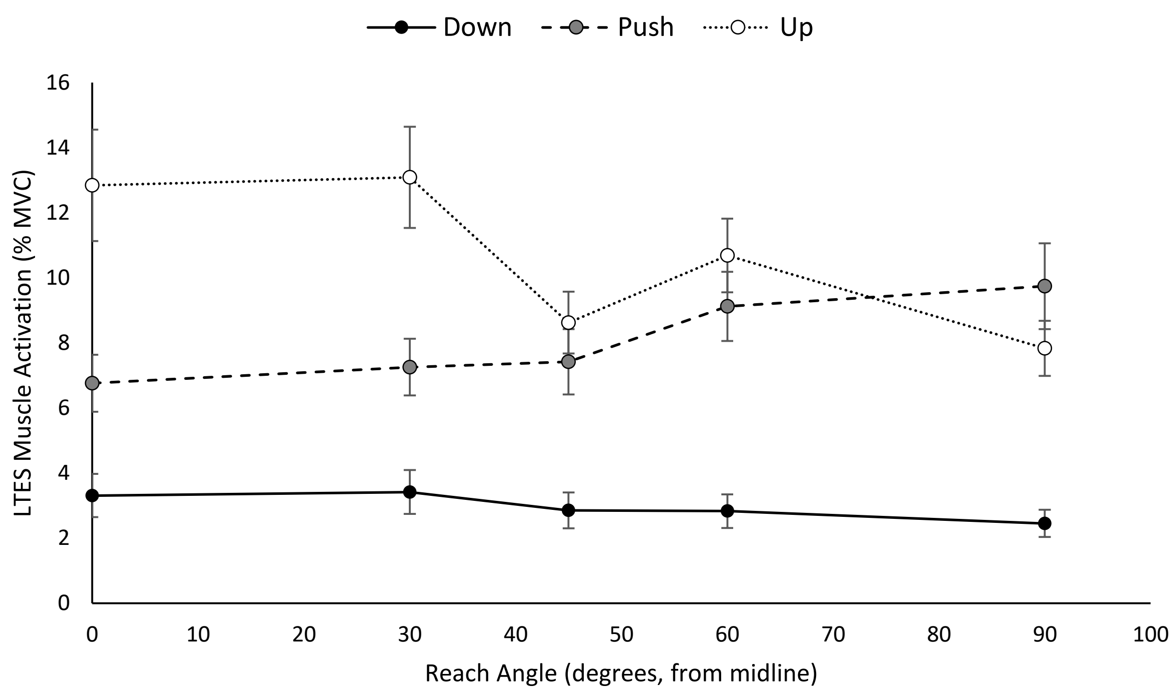


Figure 7.8: Left thoracic erector spinae (LTES) muscle activation by exertion direction and reach angle collapsed across sex, height, and load. Activation decreased with increased laterality of reach angle. Error bars represent standard error.

7.4 Discussion

Measures of thoracopelvic posture and trunk muscle activation were used to assess physical demands during simulated occupational tasks across a range of forward and lateral reach distances, task heights, and exertion directions. Thoracopelvic axial twist was used to assess low back axial twist injury risk related to current ergonomic reach distance recommendations for frequent, infrequent and occasional tasks. The physical outcomes from this study should enhance future single-handed reach envelope recommendations in the context of spine motion and loading.

Thoracopelvic (lumbar) axial twist was impacted by hand target locations and by hand force direction. As would be expected, axial twist increased with increased laterality of hand target location. Participants used a contralateral twist strategy for straight forward hand locations, with an average 7.3-

degrees of leftward lumbar axial twist across the three reach zones. This twist indicates that participants may favour lumbar twist over shoulder flexion, as this contralateral twist would reduce the effective reach distance for these hand targets. This contralateral twist strategy has been demonstrated in both standing (Cavanaugh et al., 1999) and seated applications (Reed et al., 2004). A neutral axial twist posture was achieved with hand target locations approximately 30-degrees from the midline of the body (targets B2 and C2) and then increased approximately linearly with increased angle away from the midline. The most lateral hand targets elicited approximately 15-degrees of axial twist. This magnitude of twist represents between 36 and 41% of maximum twist range of motion in an upright, neutral posture (McKinnon - Chapter 6; Drake & Callaghan, 2008; Gomez et al., 1991). A study evaluating the relationship between magnitude of vertebral axial twist and cumulative injury risk identified approximately 25% of twist range of motion as the threshold for increased relative risk and required application of a weighting-factor to increase cumulative injury risk (McKinnon - Chapter 3). This threshold applied to *in vivo* twist range of motion relates to a conservative estimate of 9-degrees. Another investigation characterized the relationship between segmental vertebral axial twist and thoracopelvic axial twist (McKinnon – Chapter 5). Using this relationship and documented maximum lumbar spine axial twist range of motion (12.0 degrees, Pearcy & Tibrewal, 1984a), this elevated injury risk would occur at 8.5 degrees of axial twist. These two methods of determining injury threshold indicate thoracopelvic axial twist greater than 8.5 degrees may pose an elevated injury risk. Previous epidemiological research has

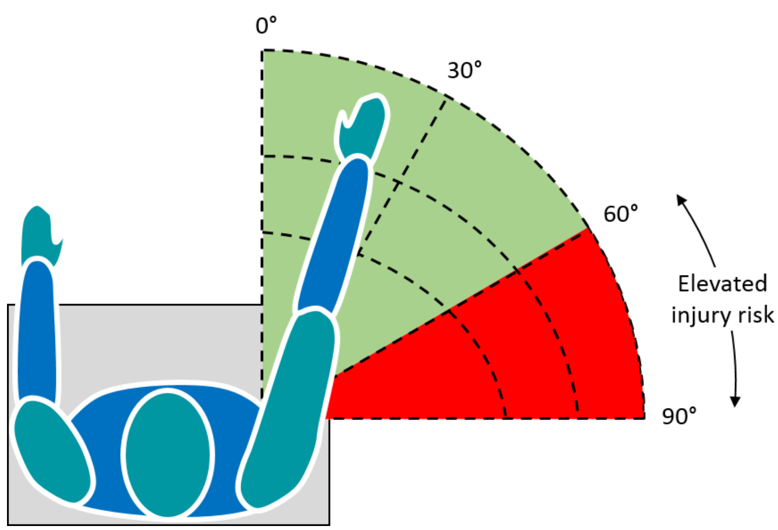


Figure 7.9: Lateral reaches greater than 60-degrees from the midline may exhibit greater injury due to lumbar axial twist and trunk muscle activation.

indicated high risk of low back injury risk with greater than 20-degrees of axial twist (Punnet *et al.*, 1991), but this work did not specifically look at twist magnitude and perhaps a lower threshold is required to identify injury risk increases. The most lateral hand target locations in the current study (A3, B4 and C4) exceed this threshold and elicit elevated injury risk regardless of task height or exertion direction (Figure 7.9). These 90-degree lateral hand target locations – targets at right shoulder width in line with the anatomical frontal plane – elicit greater axial twist induced injury risk, and future ergonomics design guidelines should suggest avoiding tasks at these locations.

Pelvis axial twist postures yielded similar patterns to thoracopelvic axial twist with contralateral twist during forward reaches and increased rightward twist with increased laterality of right side hand target location. The lower limbs and pelvis accounted for nearly 10-degrees of axial twist at the most lateral hand target locations. In the primary reach zone (A), pelvis twist was close to neutral, but increased to 4.0 degrees and 8.9 degrees of leftward twist for the directly forward hand targets in the secondary (B) and tertiary (C) reach zones, respectively. This contralateral pelvis twist is likely coupled with the same strategy seen in the thoracopelvic spine to increase effective reach distance. Lower limb kinematics were not recorded during this study, and so it is difficult to speculate on composition of lower limb postures to achieve this pelvic axial twist motion. Considering a seated posture fixes the pelvic posture and would require all twist to occur superior to the pelvis, these findings suggest that twist magnitudes and injury risk exposures at the hand locations evaluated in this study may be greater during seated work. Future investigations similar to this study are required for seated work in order to evaluate the appropriateness of current reach guidelines for seated applications.

Lumbar flexion demonstrated the opposite response of twist, with a less flexed, more upright posture at more lateral hand targets and greater flexion with increased reach distance. These results indicate a trade-off between the twist and flexion motion axes, with opposing postural demands for forward and lateral reaches. Flexion angle was similar for the primary and secondary reach zones, and showed greater flexion for the further (tertiary) reach zone. While clear trends were evident across hand target locations and exertion directions, it should be noted that lumbar flexion was less than 5 degrees across all conditions, and generally less than 3.5 degrees. The greatest flexion observed under experimental conditions represents less than 13% of flexion range of motion (Saur et al., 1996) compared to the 40% of axial twist range of motion at the most lateral hand targets. These magnitudes of flexion fall within the neutral zone (Panjabi, Duranceau, et al., 1989) and well below the elastic limit (Adams & Hutton, 1986) of the lumbar spine, and should not present an elevated injury risk during occupational reaching tasks.

Trunk extensor activation and apparent function differed by vertebral level. LES presented greatest activation bilaterally during upward exertions, with highest activation when exerting on hand targets directly forward, and higher activation with increased reach distance. Erector spinae activity is typically greater with increasingly flexed lumbar postures, as is the case with the increased reach and forward hand targets in the current study. It has been suggested that the primary role of erector spinae is spine stabilization during upright twisting (Marras & Granata, 1995; McGill, 1991; Parnianpour et al., 1988), but it must support the torso mass and has an altered line of action during combined flexion and axial twist that allows it to generate twisting torque (Marras et al., 1998). The current study confirmed these documented responses as lumbar muscle activity increased with increased forward reach distance, and elicited decreased activity with the more upright postures associated with increased laterality.

At the thoracic level, erector spinae activity appears to be related to shoulder muscle activity, posture and function. Left side TES activation and right shoulder elevation demonstrated few differences within the primary (A) and secondary (B) reach zones (activation generally less than 10% MVC), but elicited much higher activation and shoulder elevation in the tertiary (C) reach zone. These activation and elevation differences were especially evident for hand targets directly in front of the participant, and both measures decreased with increased hand target laterality. Right side TES activation was considerably higher in upward exertions (24.5% MVC) than downward (4.3% MVC) or forward exertions (10.6% MVC) regardless of target location. This likely comes as a result of an increase in the overall flexor moment on the upper spine with upward exertions. Right side activation was generally highest for directly forward hand target locations, and decreased with increasing reach distance. These findings indicate there may be a trade-off between upper back and shoulder muscle activation, as previous findings have reported shoulder muscle activation increases with increased laterality and increased reach distance beyond 300 mm (McDonald et al., 2012). There may be a transfer of load from upper back musculature at close and forward hand targets to shoulder musculature at more lateral hand targets and increased reach distances. Overall, it appears that task demands are highest for the upper back for upward exertions, which would mimic lifting or load transfer tasks.

Hip and abdominal muscles had higher demands likely related to force production and bracing strategies. Left IOB activation and right GMD presented similar trends with much greater activation during forward pushes than the other directions, and increased activation with more lateral hand locations. Haslegrave et al. (1997) reported a greater force production capability for forward pushes at lateral hand locations when using a twisted posture than when using a forward-facing posture. This is likely a driving factor for participant selection of such a twisted posture for lateral hand locations. In addition, Granata & Bennett (2005) suggested that increased muscular co-contraction and spine stiffness stabilize against external load during occupational pushing tasks. Right GMD activation exceeded 10% MVC for all forward pushes at the most lateral hand locations. With more lateral locations, participants may have shifted their weight to be predominantly supported by the right leg. This type of weight shift and rightward migration of the centre of pressure has been associated with increased GMD activation during single-leg stance (Nelson-Wong et al., 2010). Left side GMD activation was higher for forward pushes (5.5% MVC) and upward exertions (5.7% MVC) than downward exertions (2.4% MVC), but generally showed low levels of activation. It is suggested that GMD and IOB work synergistically, with the latter developing spine stiffness and the former creating hip stiffness to drive through the primary stance leg and generate distal hand force. From a task demand perspective, right GMD and bilateral IOB muscle activation exceeded 10 %MVC for forward pushes at all hand target locations greater than 60-degrees from the body midline, regardless of reach distance. Future ergonomics design guidelines should suggest all pushing tasks remain in front of the body, rather than at lateral hand target locations.

One of the objectives of this study was to enhance single-handed reach envelope guidelines in the context of spine motion and loading. Thoracopelvic axial twist was characterized across a range of hand target locations, and indicates that future ergonomics guidelines should suggest limiting design of reaching tasks beyond 60-degrees from the midline of the participant or worker. Beyond this threshold, lumbar intervertebral injury risk may be elevated, and hip and abdominal muscles had increased activation. It appears that while shoulder demands are appropriate for derivation of forward reach guidelines and task recommendations, lumbar axial twist is the primary concern for lateral reaching tasks, and future guidelines should be adjusted accordingly.

7.5 Conclusion

The results of this study characterized thoracopelvic posture and motion related to hand target location for right-handed, single-handed exertions. As expected, a combination of lumbar axial twist and pelvic twist increased with increased laterality of hand location. The magnitude of axial twist at the most lateral hand locations (90-degrees from body midline), was suggested to present elevated risk for intervertebral injury, regardless of the task zone from frequent to occasional. Participants generally flex more with increased forward reach distance, and stood more upright as tasks became more lateral. Hip and abdominal muscle activation exceeded 10% MVC for these very lateral tasks, and had the highest activation for forward push exertions. These findings suggest that future ergonomics guidelines should assess reaching and exertion tasks from hand target locations beyond 60-degrees, from the midline of the body and consider them as non-optimal zones.

CHAPTER 8
GENERAL DISCUSSION

8.1 Thesis Questions Revisited

The global objective of this thesis was to investigate the relationship between working postures used in industrial tasks and lumbar axial twist injury risk. The following is a summary of these specific thesis findings with respect to this global objective and the thesis questions posed in Chapter 1 (Section 1.6.1).

What is the relationship between externally measured thoracopelvic axial twist and the actual segmental axial twist motion of the intervertebral joints?

Study 2 (Chapter 5) demonstrated there is a linear increase in vertebral axial twist with thoracopelvic axial twist at each lumbar vertebral level. The largest amount of motion was at the L2-L3 (46.8% of lumbar motion) and L5-S1 (33.5%) intervertebral joints, and this motion was characterized by a linear equation for each vertebral level. These findings suggest that L4-L5 may not be an appropriate site to represent the low back in biomechanical modelling approaches. The aggregate lumbar motion accounted for just over 30% of the total thoracopelvic range of motion, which is similar to other studies using MRI and CT images of human participants.

Can we use ultrasound as a modality to consistently and accurately measure vertebral axial twist motion?

In Study 1 (Chapter 4), ultrasound measures of vertebral axial twist were directly compared to measurements from a motion capture system using a porcine function spinal unit. Agreement between the two systems was excellent, with an overall correlation coefficient of 0.90, and 0.014 degrees of mean error between the two systems. The agreement between the systems indicates that ultrasound can produce valid measures of vertebral axial twist, and low intra-rater error for both *in vitro* and *in vivo* studies providing there are consistent identification of clear landmarks on the ultrasound images.

What amount of lumbar axial twist presents an elevated injury risk for working populations? Using a porcine cervical model of the human lumbar spine, Study 3 (Chapter 6) characterized an exponential relationship between absolute and relative magnitude of axial twist motion and cumulative axial twist moment failure tolerance. A threshold of approximately 25% of intervertebral motion was identified and used to construct weighting-factor functions to improve estimates of injury risk related to cumulative load in twisting. When combined with the relationship between external and internal twist motion developed in Study 2 (Chapter 5), it appears that approximately 8.5 degrees of lumbar axial twist presents an elevated level of cumulative injury risk for working populations.

What movement strategies do people use to perform reaching tasks at different hand locations, and how do task parameters impact these strategies?

Study 4 (Chapter 7) characterized trunk and right upper limb posture for reaching exertions at hand locations corresponding to current ergonomic reach design guidelines. Participants tended to flex their lumbar spine for directly forward reaches, and then exhibit lumbar axial twist and stand more upright for lateral reaches. The most lateral reaches elicited 15 degrees of trunk axial twist (40% of ROM), regardless of the task zone (frequent, infrequent, or occasional). Pushing exertions require more lateral body positioning to produce the required force and caused an increase in both lower limb and thoracopelvic twisting. Muscle activation followed the same principals, with the upper back musculature being most active for forward reaches, and the abdominal and hip musculature most active during twisted, pushing exertions.

8.2 Major Contributions

***Contribution #1*: Future ergonomics reaching guidelines should aim to remove tasks beyond 60-degrees from the midline of the body.**

One of the primary aims of this thesis was to improve current ergonomics reach guidelines and provide a physiological and biomechanical basis for reach distance recommendations incorporating the low back. The results from Studies 2, 3 and 4 (Chapters 5, 6 and 7, respectively) link together to identify a threshold for increased axial twist cumulative injury potential using measurement of vertebral twist and participant-selected postures to perform manual exertions at hand targets related to current reach guidelines. Study 3 (Chapter 6) provided a specific threshold for intervertebral injury risk, Study 4 (Chapter 7) identified posture strategies used for different hand target locations, and Study 2 (Chapter 5) developed the link to the vertebral motion in these selected postures, and allows application of the injury threshold at a thoracopelvic posture level. Specific development of a recommended axial twist design guideline is outlined in Section 8.3.

***Contribution #2*: There is a linear motion pattern for both internal vertebral and external thoracopelvic axial twist motion across the full range of motion.**

Few studies have characterized axial twist across the full range of motion at either the external or vertebral joint level. Typical medical imaging investigations have quantified vertebral posture at a neutral and twisted position with no identification of the motion profile between those postures (Haughton et al., 2002; Ochia et al., 2006; Pearcy & Tibrewal, 1984a). Study 2 (Chapter 5) reported that vertebral and intervertebral axial twist increase linearly at all lumbar levels with increased thoracopelvic twist magnitude. The correlation on these linear fits ranged from 0.43 to 0.79 across vertebral levels. This linear fit simplifies the relationship between internal and external motion, and indicates that non-linearity in vertebral twist may result from issues in the surrounding structures. During simulated industrial exertion tasks, Study 4 (Chapter 7) demonstrated a linear increase in thoracopelvic axial twist with increasing angular hand target location. This simplified relationship justifies the use of an angular threshold in future ergonomics design guidelines for lateral reaching tasks.

***Contribution #3*: Ultrasound is a valid modality for measurement of in vivo axial twist in the lumbar spine.**

External measurement of intervertebral motion is hampered by skin motion artifact issues (Heneghan & Balanos, 2010; Mörl & Blickhan, 2006) or invasive and impractical procedures (Steffen et al., 1997). Alternatively, medical imaging modalities such as MRI and CT are able to accurately measure segmental twist, but present major challenges for task-oriented upright postures and dynamic investigations, as they require a participant to be in a fixed posture within a confined imaging space. Study 1 (Chapter 4) validated use of an ultrasound imaging technique to measure axial twist, and Study 2 (Chapter 5) was able to use this technique to measure *in vivo* vertebral axial twist in relation to externally measured thoracopelvic twist. The portability and cost-effectiveness of an ultrasound system along with the accuracy in these studies identifies ultrasound as a feasible modality for measurement of axial twist during occupational, task-specific movements. Segmental twist can be measured under controlled external thoracic axial twist conditions and in response to mechanical loading for future investigations of lumbar spine twist injury mechanisms.

Specific Study Contributions

Each study of this thesis was essential for development of a simple, cost-effective technique for in vivo measurement of lumbar spine axial twist, and to develop suggested improvements to current ergonomics guidelines that could reduce injury-causing lumbar axial twist postures. Study 1 (Chapter 4) developed and validated an ultrasound technique to measure intervertebral twist motion, and Study 2 (Chapter 5) used that technique to directly link gross thoracopelvic motion to individual intervertebral joint motion. Study 3 (Chapter 6) developed an individual intervertebral joint model based on axial twist magnitude and gave an estimate for cumulative twist injury, and Study 4 (Chapter 7) directly measured working postures and loading strategies during simulated work. These follow the global thesis structure outlined in Figure 1.4, with the technique and outputs from Studies 1 and 2 creating a direct link between intervertebral injury risk defined in Study 3 and the working postures described in Study 4. The links between these individual studies are described in detail in the following section (8.3), and Figure 8.1 presents the study outputs in parallel with the structure of Figure 1.4.

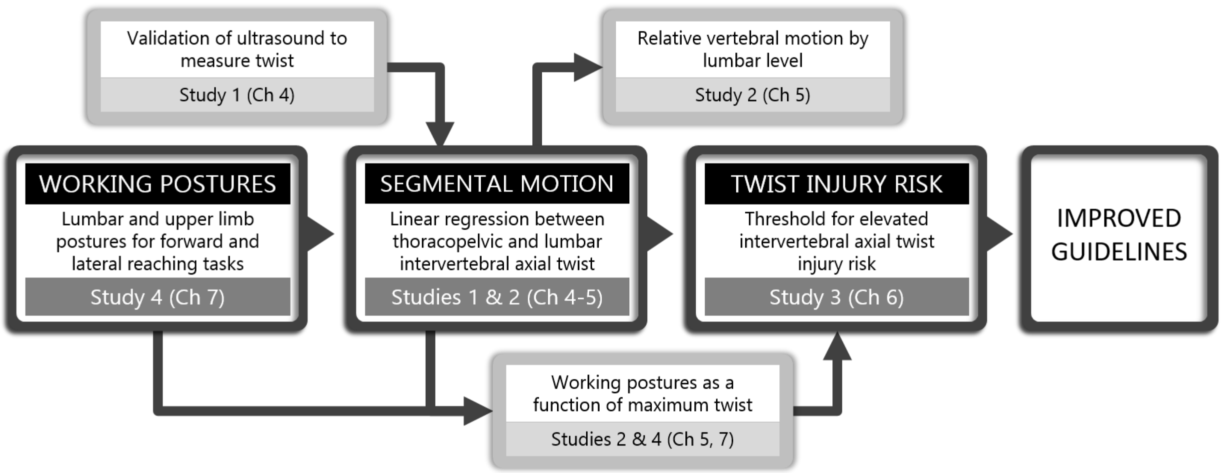


Figure 8.1: Specific study outcomes from the global thesis structure perspective.

8.3 Suggestions for Future Ergonomics Reach Guidelines

Current guidelines seek to apply ergonomics standards to the development, design, use, management, and improvement of work systems. The ultimate goal is to enable an organization to enhance worker health, safety, and well-being and to optimize system performance to prevent occupational injuries related to occupational activities and work environments. Ergonomics has its greatest benefit when used early in the work design process, as it enables engineers and designers to integrate human factors into system design before workers are ever exposed to loads. Tools such as reach envelope guidelines possess easy implementation and ensure worker capabilities are integrated into workstation designs. While built on sound principles, current guidelines lag behind modern research and primarily draw on material more than 15 years old. Current guidelines include recommendations for forward and lateral reach distances based on task frequency and anthropometry (CSA, 2012). Forward reach guidelines focus on shoulder posture and mechanics which attempt to minimize load moments, fatigue and perceived exertion (Anton et al., 2001; Brookham et al., 2010; Dickerson et al., 2006). Lateral reach guidelines are underdeveloped and generally apply these forward reach concepts to the lateral reach envelope. With a specific focus on reach guidelines, this thesis sought to identify biomechanical factors to develop improved recommendations for single-handed lateral reaching task distances and hand target locations.

To develop improved guidelines, this thesis linked a threshold for increased axial twist injury risk (Study 3) to working postures for multiple hand target locations (Study 4) using a relationship between thoracopelvic and vertebral posture (Study 2):

1. Injury risk is elevated at 22.7% of intervertebral axial twist range of motion.
2. Maximum lumbar intervertebral motion occurs at L2-L3, accounting for 46.8% of lumbar motion:

   $PROP_{L2\text{-}L3}$ = 0.468
3. Given a documented lumbar spine twist range of motion of 12.0 degrees (Pearcy & Tibrewal, 1984a), estimated maximum L2-L3 motion is 5.6 degrees, and the associated elevated injury risk threshold would be 1.3 degrees:

   $$\begin{aligned} ROM_{LAT} &= 12.0 \text{ degrees} \\ ROM_{L2\text{-}L3} &= ROM_{LAT} * PROP_{L2\text{-}L3} * 0.227 \\ &= 1.3 \text{ degrees} \end{aligned}$$

   where ROM_{LAT} is lumbar axial twist (VAT) range of motion, and $ROM_{L2\text{-}L3}$ is L2-L3 range of motion

4. With the slope (m) of the linear relationship between thoracopelvic posture and vertebral axial twist developed in Study 2, this elevated risk would correspond with 21.2% of thoracopelvic range of motion, or 8.7 degrees of thoracopelvic twist based on a 40-degree maximum range of motion.

m_{L2} = 0.174
m_{L3} = 0.114
$RISK_{L2\text{-}L3} = ROM_{L2\text{-}L3} * (m_{L2} - m_{L3})$
= 21.2% ROM

ROM_{TAT} = 41.1 degrees
$RISK_{TAT} = ROM_{TAT} * RISK_{L2\text{-}L3}$
= 8.7 degrees

where $RISK_{L2\text{-}L3}$ is the percentage of thoracopelvic range of motion associated with elevated cumulative injury risk, and $RISK_{TAT}$ is the thoracopelvic axial twist (TAT) threshold for this elevated risk

Hand targets with an angular position 60-degrees from the midline of the body were associated with a mean thoracopelvic axial twist angle of 7.2 degrees collapsed across all conditions, while hand targets at 90-degrees had 14.7 degrees of twist (Study 4). Therefore, all hand targets less than 60-degrees fell under the threshold, $RISK_{TAT}$, for elevated cumulative injury risk, while hand targets at 90-degrees exceeded this threshold. These elevated risk calculations suggest that future ergonomics design guidelines should maintain work reach dimensions for primary, secondary, and tertiary zones, but reaching tasks to hand target locations greater than 60-degrees from the midline of the body should be avoided. Figure 8.1 illustrates these recommended changes mapped onto current reach envelope guidelines.

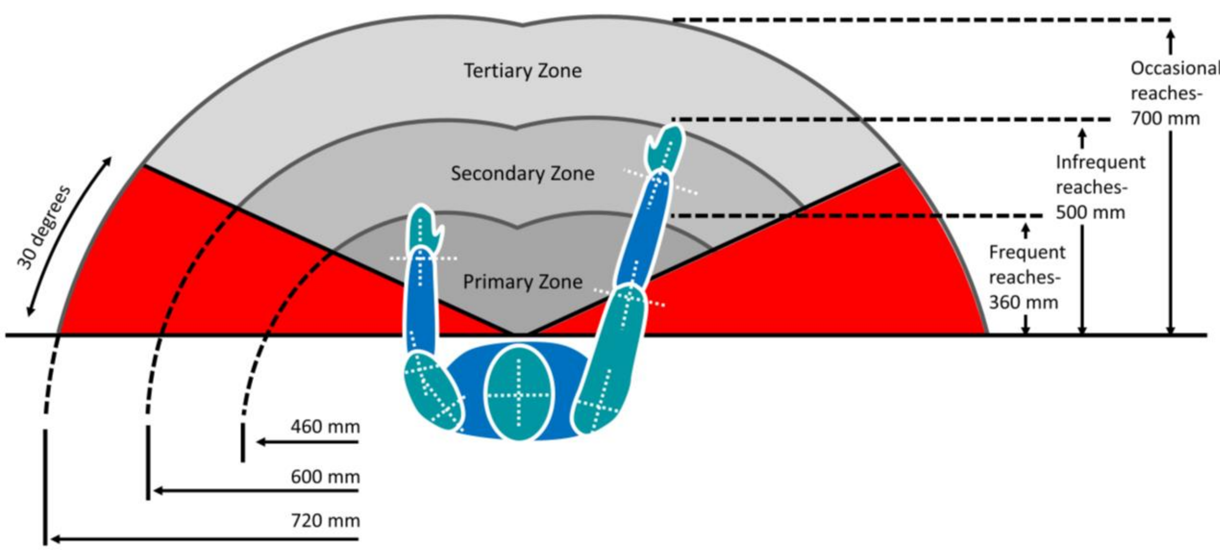


Figure 8.2: Recommended reach envelope updates mapped on to current guidelines. Secondary Reaches beyond 60 degrees from the midline of the body should be avoided to limit thoracopelvic axial twist and an associated elevated cumulative injury risk. Figure adapted from CSA (2012).

8.4 General Limitations

The updated reach envelope recommendations are dependent on the relationship between a porcine FSU failure model and human working postures. Because of the nature of acquiring and measuring human spine motion, functional range of motion and true range of motion are often confused and present a potential source of error in these recommendations. Studies evaluating vertebral range of motion using radiographs or Steinmann pins have found range of motion to be between 2-5 degrees (Gunzburg et al., 1991; Pearcy & Tibrewal, 1984a). While perhaps more functionally accurate, these studies may underestimate the true range of motion of an isolated intervertebral joint. In a study more closely resembling the setup used in Study 3, Farfan et al. (1970) found cadaveric range of motion to be 22.6 degrees. This is similar to the 20.6 degree range observed for porcine specimens from a neutral posture (Drake, 2008), but it is suggested that relative motion, rather than absolute measures, should be used when applying the weighting-factors and elevated risk thresholds from Study 3 for use in developing updated reach guidelines.

The relationship between thoracopelvic and vertebral axial twist postures from Study 2 is only applicable for isolated axial twist motions. Twist range of motion has been consistently measured between 37-41 degrees from an upright, neutral posture (McKinnon - Chapter 2; Drake & Callaghan, 2008; Gomez et al., 1991), but is reduced to 30.1 degrees when coupled with extension and increased to 43.9 degrees with flexion (Drake & Callaghan, 2008). This altered range of motion may impact the relationship between external and internal measures, and may make the calculated linear regression model inappropriate for motions coupled with large magnitudes of flexion. Physical demands on the participant in a fixed, constrained posture made recording of vertebral axial twist in flexed and extended postures a major challenge, and required a streamlined experimental protocol. Future studies will need to modify the testing apparatus to make such recordings feasible and safe for participants.

While ergonomics recommendations are intended to be applied to many setups and work conditions, the research that drives such guidelines is often specific and controlled. The working postures and subsequent reach envelope recommendations that emerged from Study 4 are specific to single-handed, upright standing exertions, and cannot be universally applied to any reaching task. For example, a two-handed reaching task may require even more conservative guidelines, as total upper body mobility would be limited by the contralateral reaching arm. Similarly, seated exertions would not necessarily align with the postures and goals from this thesis, as pelvis and lower limb mobility would be constrained. While extensive work has tied seated exertions to operator reach envelopes (Faulkner & Day, 1970; Godwin et al., 2007; Reed et al., 2004), standing two-handed exertions have not been fully characterized and require future research to further improve upon current reach envelope guidelines.

CHAPTER 9

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