The effect of limited hip mobility on the lumbar spine in a young adult population

by

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Author's 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.

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Abstract

Limited hip mobility is known to affect the lumbar spine. Much of the previous research has utilized a participant population whose hip mobility is compromised due to arthritic or neurological dysfunctions. Such aetiologies may confound the outcomes, as their effects may not be limited to the hip. The purpose of this thesis was to recruit a healthy young adult population with limited hip mobility to further investigate its effect on the lumbar spine, as well as the role of exercise intervention.

Several cascading studies were conducted that were unified around a central theme of links between hip and spine function:

Study # 1 investigated the normal distribution of passive hip extension and rotation in a group of 77 males (age 19-30). Data was collected using an infra-red motion capture system and compared to goniometric measurements. The resulting angles represent the 5^{th} – 95^{th} percentiles, including the averages and standard deviations.

Study # 2 compared movement patterns between groups of males with limited and excessive hip mobility. Participants were required to perform simple functional activities (lunging, twisting, walking, etc) as well as use the elliptical trainer. Resulting hip and spine angles demonstrated that the men with limited hip mobility stood with a more anteriorly tilted pelvis, and assumed a posture with more lumbar and hip flexion on the elliptical trainer, compared to those with greater mobility. This, in turn, resulted in a greater lumbar compression load due to increased back muscle activity.

Study #3 involved recruitment of 24 young adult males with limited hip mobility. Their movement patterns were assessed (as in study #2), then they were assigned to one of four intervention groups: hip stretching, spine stabilizing, hip stretching combined with spine

stabilization, and control. Participants in the 3 exercise groups attended supervised exercise sessions once/week for 6 weeks, but were expected to exercise a minimum of 4 times/week on their own. At the end of the 6 weeks, intake parameters were re-assessed, and movement pattern assessment repeated. Despite significant increases in available hip flexibility and/or large increases in trunk muscle endurance and trunk motor control, there were few indications that participants were any more adept at decreasing lumbar motion, or utilizing their newfound hip flexibility during functional activities.

Study #4 compared those in the 10^{th} and 90^{th} percentiles of available hip rotation, using a frictionless apparatus to investigate passive stiffness properties of the hip. Participants adopted a posture of upright standing, with one leg supported on a turntable apparatus, and upper body and pelvis secured. A an applied rotational moment resulted in passive hip internal and external rotation.   Outcomes demonstrate that those with limited hip mobility stand with the leg more externally rotated and require a larger moment to initiate motion. Passive stiffness curves indicate greater stiffness properties in those with limited hip mobility, and more resistance to an external rotation moment than internal rotation.

Study #5 investigated passive hip stiffness in the sagittal plane, comparing those with limited and excessive hip extension. Using a frictionless jig, with the participants lying on their left side, the left hip was pulled into extension with knee position varying. Those with limited hip mobility demonstrated increased passive stiffness compared to the more mobile group, and stiffness was greater when the knee was in extension. The group with limited mobility also showed a trend of increased back extension compared to the more mobile group, when the hip and lumbar spine were both free to react to the applied extension moment.

Study #6 summarizes the spine/hip kinematics and muscle activation levels produced when using the elliptical trainer, as well as lumbar compressive and shear forces. It differs significantly from walking in that it produces more lumbar motion in flexion/extension and lumbar twist, but less lateral bend. Participants also tended to adopt a greater mean lumbar flexion angle on the elliptical, which in turn resulted in greater muscle activity in the back extensors. Varying hand position, velocity and stride length were all found to significantly affect the amount of lumbar motion. Highly phasic muscle activity is seen, with the gluteal muscles and internal obliques demonstrating the greatest activation levels.

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Finally, a huge thank-you to my children: Geoff, Christine and Tom, who never questioned why their mother felt the need to attend university at the same time as them. You grounded me, and were a constant reminder as to what really matters in life.

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List of Abbreviations

Muscles:

BF: biceps femoris

EO: external oblique

ES: erector spinae

GMax: gluteus maximus

GMed: gluteus medius

GMin: gluteus minimus

HS: hamstrings

IO: internal oblique

LD: latissimus dorsi

Mult: multifidus

PD: posterior deltoid

PM: pectoralis major

RA: rectus abdominis

RF: rectus femoris

SM: semimembranosis

ST: semitendinosis

TFL: tensor fasciae latae

Other:

AIIS: anterior inferior iliac spine

A/P: anterior/posterior

ASIS: anterior superior iliac spine

BP: blood pressure

BW: body weight

cpm: cycles per minute

COM: centre of mass

EHE: excessive hip extension

EHM: excessive hip mobility

EHR: excessive hip rotation

EMG: electromyography

Horiz: horizontal

Lt: left

LHE: limited hip extension

LHM: limited hip mobility

LHR: limited hip rotation

LSB: left side bridge

LBP: low back pain

MVC: maximum voluntary contraction

MTT: modified Thomas test

OA: osteoarthritis

PSIS: posterior superior iliac spine

Rt: right

Sag: sagittal

SD: standard deviation

Med/Lat: medial/lateral

1. General Introduction

There exists a close kinematic relationship between the hips and the lumbar spine. Limited hip extension and/or rotation are known to affect lumbar spine position and motion, and have been implicated as one of the precursors to low back pain (Barbee Ellison, Rose, & Sahrmann, 1990; Cibulka, Sinacore, Cromer, & Delitto, 1998; Cibulka, 1999; L. W. Lee, Kerrigan, & Della Croce, 1997; Mellin, 1988; Mellin, 1990; Offierski & Macnab, 1983; Schache, Blanch, & Murphy, 2000; Sjolie, 2004; Thurston, 1985; Vad, Gebeh, Dines, Altchek, & Norris, 2003; Vad et al., 2004). Most of this research, however, has been carried out on populations that have limited hip motion due to neurological or arthritic dysfunctions, or who are classified in a low back pain population. Outcomes from such groups may be confounded by movement pattern dysfunctions in the spine that are similar to those demonstrated in the hip (i.e. limited movement or altered reflex control). Ideally, studying a group with limited hip mobility, but with no overlying neurological, arthritic or painful conditions might facilitate a clearer understanding of how hip mobility affects the lumbar spine in a healthy population. Clinically, there exists such a group of pain-free young adults: they demonstrate limited hip mobility of unknown etiology, but with no known neurological or arthritic changes. This thesis was prepared as a series of studies which focus on this group of healthy young adults, to further the understanding of how hip mobility, and lack thereof, affects motion and mechanics in the lumbar spine.

Normally, relaxed standing results in the pelvis rotating slightly posteriorly, such that the strong ilio-femoral ligaments on the anterior hip joints resist further rotation, and erect posture can be maintained without active muscular control (Aspden, Rudman, & Meakin, 2006; Fuss & Bacher, 1991). Increased anterior pelvic rotation will increase the sagittal angle of inclination of the pelvis, which typically sits at approximately 30 - 32°, calculated by drawing a line along the top of the sacrum (Magee, 1987; S. M. McGill & Norman, 1986) or approximately 5.4° - 12° if calculated as the angle between the two posterior superior iliac spines (PSISs) and the anterior superior iliac spines (ASISs) (Schache, Blanch, Rath, Wrigley, & Bennell, 2003; Smidt, McQuade, Wei, & Barakatt, 1995; Whittle & Levine, 1995) . The plateau on the top of the sacrum forms the base for the L5 vertebrae. Consequently, changing the amount of anterior or posterior pelvic tilt will also affect the sagittal angle of the lower lumbar spine (Crosbie, Vachalathiti, & Smith, 1997b; Dunk, Kedgley, Jenkyn, & Callaghan, 2009; Whittle & Levine, 1995).

The question arises as to how this carefully balanced system reacts when the structures around the hip joint impose restrictions due to decreased length and/or have an altered length/tension relationship. What happens to the position of the pelvis and lumbar spine if the anterior hip structures (passive or active) are tighter than normal? Similarly, where does the normal rotation associated with activity occur if hip joint rotation is compromised? While recognizing that most of the studies addressing these questions have used an arthritic or neurologically-impaired population, the literature suggests that lack of hip extension motion results in an increased anterior tilt of the pelvis, as well as increased total range of sagittal pelvic motion (Kerrigan, Lee, Collins, Riley, & Lipsitz, 2001; Kerrigan, Xenopoulos-Oddsson, Sullivan, Lelas, & Riley, 2003; L. W. Lee et al., 1997; Perry, 1992; Schache et al., 2000; Thurston, 1985). Although there appears to be little information regarding the effect of limited hip rotation on pelvic and spine kinematics, there are many publications linking lack of hip joint rotation with low back pain (Barbee Ellison et al., 1990; Cibulka et al., 1998;

Fairbank, Pynsent, Van Poortvliet, & Phillips, 1984; Scholtes, Gombatto, & Van Dillen, 2009; Vad et al., 2003; Vad et al., 2004; Van Dillen, Bloom, Gombatto, & Susco, 2008).

Studying the effect of limited hip mobility on the lumbar spine in a young healthy population decreases the likelihood that the outcomes will be confounded with associated changes in the spine due to arthritis, neurological dysfunctions, or pain-avoidance. Anecdotally, the majority of these young adults tend to be athletic with a high fitness level, thus perhaps increased muscle stiffness secondary to hypertrophy is a factor. The lack of mobility could also be due to habitual postures and activities, such as those seen in cycling and ice hockey. Such activities require the hips to be in a flexed position for most of the sport-related time, with little opportunity for active hip extension, which would tend to intermittently stretch the anterior structures into a lengthened position. This tendency for less mobility in an athletic population has been described by Manning and Hudson (2009) who found limited hip mobility in 2 groups of male soccer players: youths (age 16 – 18 yrs.) and senior (age 17 and up, mean 26.3(4) yrs), when compared to age-matched non-athletic males. Hip internal rotation (IR) and the Faber's test (a combination of flexion/abduction and ER described by Ross et al (2003)) were both significantly less in the soccer players ($p < 0.001$ in both measurements).

It was felt that studying a young, healthy group would give additional insight into compensatory mechanisms that occur in the pelvis and spine when hip mobility is compromised. The resulting outcomes would compliment previous studies in the literature, which have tended to focus on an older group, or one with abnormal motor patterns. Consequently, the motivation behind this thesis was to address several related questions, as described below.

1.1. Global Thesis questions

1.1.1. What defines limited hip mobility in a young adult male population?

The literature is mixed in defining normal range of motion (ROM) in this population group. Results differ based on the technique of measurement, for example: active vs. passive, position (supine vs. prone), and technique (modified Thomas test vs. prone extension). Thus, the first data collection explored normal hip ROM in a young adult male population, using positions and methods relevant to a clinical setting, but capturing the position data using a state of the art motion capture system. Normative data was collected over a large number of males, age 19 – 30, measuring hip extension and rotation. From this, percentile data was calculated, including the mean and standard deviation (SD), thus helping to understand what could be considered normal, limited and excessive hip mobility.

1.1.2. Do young adult males with limited hip mobility move differently than those with normal or excessive ROM?

This is a seminal question. Having defined what constitutes limited and excessive hip mobility, two groups were formed to take part in numerous active movement trials. Activities were chosen that represent common exercises or activities requiring moderate amounts of hip extension and/or rotation. Data was collected on hip and spine motion, in an attempt to understand the effects of limited hip mobility on lumbar spine motion. In a separate data collection, the response of passive tissues in these two groups was compared. Using a relatively frictionless measurement restraint jig, an extension or rotation moment was applied to the distal end of the leg, while motion of the hip and/or back were monitored, and relative passive resistance to motion at the two segments was quantified.

1.1.3. *In a group of young adult males with limited hip mobility, is it possible to enhance their hip range of motion and/or core strength and endurance with a 6 week exercise protocol? Do such gains subsequently result in changes to movement patterns in the hip and back?*

Based on a preliminary assumption that the group with limited mobility will tend to extend and/or rotate more in their spine than the looser group, the question arises, "Can we change how they move?" Men with limited hip mobility participated in the same movement trials discussed in section *1.1.2.* They were randomly assigned to 4 different exercise groups, aimed at increasing their hip mobility and/or improving the strength and motor control of their spine stabilization muscles. At the end of the 6 weeks, the movement trials were repeated, and hip ROM and spine stabilization strength re-assessed. The outcomes from this study addressed the following specific questions:

- To what extent does hip mobility change in this group of young men with limited mobility with a 6 week protocol?
- Does spine stabilization muscle strength and/or motor control improve with a 6 week treatment protocol?
- Assuming that strength and mobility do improve, does this make any difference to how a person moves?

1.1.4. *How do the passive stiffness qualities differ in groups defined as having limited or excessive hip mobility?*

Two separate data collections focused on the passive resistance to an applied moment at the hip joint: one in upright standing with an applied hip rotation moment, and a second in side lying with an applied extension moment. Passive stiffness properties between the two groups (those with limited mobility, and those with excessive mobility) were compared, as well as the changes that occur when the knee is flexed or extended in the extension trial.

1.2. Data collections

Although there were 6 studies that emerged in this thesis, all data was obtained using 4 data collections, which are explained as follows:

1. *Normative data*

Passive hip extension and rotation measurements were obtained in a group of young males, age 19 – 30. Position data was captured using the Vicon MX Motion system (Vicon Motion Systems, Oxford, UK), and subsequently used to calculate joint angles. The number of participants was determined by how many were required to ensure a normal distribution. Percentile data was then calculated, indicating the hip mobility measurements that constitute the various percentiles.

2. *Movement/motion trials*.

This phase of collection had several parts: firstly, males with limited hip mobility were compared to those with excessive mobility in a number of movement trials. They were asked to walk, twist, lunge, perform active hip extension (in standing), and exercise on the elliptical trainer. Full body linkage movement was recorded via the use of reflective markers and the Vicon system, while trunk muscle activation levels were collected using surface electromyography (EMG). Joint angles and kinetic data were then calculated and compared between the two groups.

Subsequently, participants with limited hip mobility underwent a 6 week training intervention program, being randomly assigned to one of 4 exercise groups:

- Hip stretching

- Spine stabilization
- Hip stretching and spine stabilization
- Control (no intervention)

At the end of the 6 weeks, they underwent the same movement assessment protocol described previously. Comparisons were made as to differences that occurred in flexibility, strength/endurance, and movement patterns of the hip and spine.

3. *Passive rotation*

Passive stiffness properties of the hip were explored using a frictionless jig apparatus. Those with limited hip rotation were compared to those with excessive rotation. Participants were secured in two legged upright standing, but with the leg of interest standing on a turntable apparatus. A rotation moment was applied to the apparatus, passively pulling the hip into IR or ER. From these data, passive stiffness properties were calculated, as well as information about the preferred resting position of the limb.

4. *Passive extension*

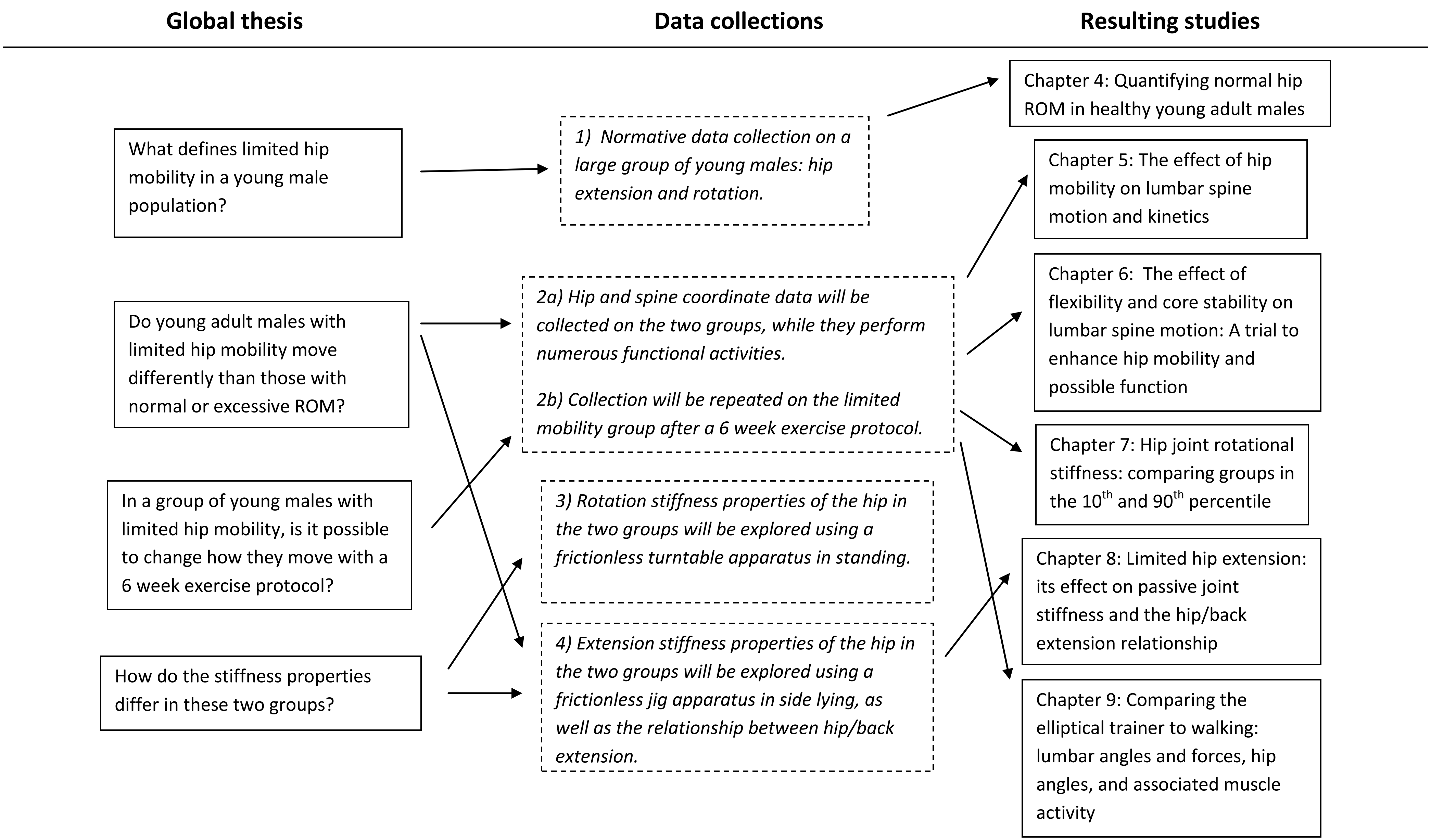
The passive stiffness properties of hip extension were also studied, comparing those with limited and excessive hip extension. This took place in a frictionless jig apparatus which required participants to lie on their side, with the limb of interest and pelvis securely fastened to 2 mobile platforms, the trunk and upper body to an immobile one. An extension moment was applied to the distal apparatus, with variations in knee flexion position, to investigate how hip extension differed between these two groups and positions. The temporal relationship between hip and back extension was explored, as well as the preferred position of elastic equilibrium between the two mobility groups.

1.3. Resulting studies

This document includes 6 separate studies which are the result of 4 data collections, and will be written in a "manuscript" format. That is, each study will be written as it would be for submission to a peer reviewed journal, independent of each other study. Consequently, there may be some repetition in the introduction and literature reviews of each study. The 6 studies are as follows:

1. Chapter 4: *Quantifying normal hip ROM in healthy young adult males*
2. Chapter 5: *The effect of hip mobility on lumbar spine motion and kinetics*
3. Chapter 6: *The effect of hip flexibility and core stability on lumbar spine motion: A trial to enhance hip mobility and possible function*
4. Chapter 7: *Hip joint rotational stiffness: comparing groups in the 10^{th} and 90^{th} percentile*
5. Chapter 8: *Limited hip extension: its effect on passive joint stiffness and the hip/back extension relationship*
6. Chapter 9: *Comparing the elliptical trainer to walking: lumbar angles and forces, hip angles, and associated muscle activity*

The relationships between the original questions, data collections and individual studies are outlined on the following page.


Global thesis
Data collections
Resulting studies
What defines limited hip mobility in a young male population?
Do young adult males with limited hip mobility move differently than those with normal or excessive ROM?
In a group of young males with limited hip mobility, is it possible to change how they move with a 6 week exercise protocol?
How do the stiffness properties differ in these two groups?
1) Normative data collection on a large group of young males: hip extension and rotation.
2a) Hip and spine coordinate data will be collected on the two groups, while they perform numerous functional activities.
2b) Collection will be repeated on the limited mobility group after a 6 week exercise protocol.
3) Rotation stiffness properties of the hip in the two groups will be explored using a frictionless turntable apparatus in standing.
4) Extension stiffness properties of the hip in the two groups will be explored using a frictionless jig apparatus in side lying, as well as the relationship between hip/back extension.
Chapter 4: Quantifying normal hip ROM in healthy young adult males
Chapter 5: The effect of hip mobility on lumbar spine motion and kinetics
Chapter 6: The effect of flexibility and core stability on lumbar spine motion: A trial to enhance hip mobility and possible function
Chapter 7: Hip joint rotational stiffness: comparing groups in the 10th and 90th percentile
Chapter 8: Limited hip extension: its effect on passive joint stiffness and the hip/back extension relationship
Chapter 9: Comparing the elliptical trainer to walking: lumbar angles and forces, hip angles, and associated muscle activity

2. Common Review of the Literature

2.1. Applied Anatomy

In order to understand how the hip and back interact during different activities, a thorough knowledge of the available literature on mobility is required. Thus, the applied anatomy of the hip, pelvis, and lumbar spine will be discussed in detail with regards to normal range of motion. Static mobility will be discussed in this section, followed by a subsequent description of movement associated with gait.

2.1.1. *Hip joint*

The hip joint is a multi-axial ball and socket joint, allowing the femur to rotate about 3 orthogonal axes within the acetabulum of the pelvis (or, conversely, allowing the acetabulum to rotate around the femoral head). The translation motions associated with rotation are seldom mentioned in the literature, and are generally considered to be insignificant in comparison with the rotations. A combination of a deep bony socket, additional acetabular labrum, strong capsule, ligaments and muscles surrounding the joint all result in great stability (Aspden, Rudman, & Meakin, 2007; Magee, 1987).

Clinical assessment of the hip joint includes objective measurements of the femoral motion in the 3 orthogonal planes: flexion/extension (sagittal plane), abduction/adduction (frontal plane), and internal/external rotation (horizontal plane). Usually, the first 4 measurements are recorded in the supine position, allowing both visual and manual monitoring of unwanted pelvis motion. Internal and external rotation are more accurately measured in the prone or sitting position, with the knee flexed to 90°. However, changes in the sagittal plane orientation of the joint with these two positions (supine vs. prone) may alter the tension in the soft tissues surrounding the hip, resulting in different measurement outcomes between the two. Consequently, consistency with the position chosen and documentation of the same is important when measuring hip rotation.

Table 2-1 demonstrates the variability of “normal” hip motion, as described in the literature. As can be seen, there exist many discrepancies as to what is considered normal range of motion (ROM), which may be due to differences in sex, position and age of the participants.

Table 2-1: Normal range of hip joint ROM in degrees, as published in the literature. Dashed line indicates information was not supplied.

Authors	**Age group**	**N**	**sex**	**flex**	**ext**	**abd**	**add**	**ER**	IR
Boone et al, (1978)	18-19	53	m	123(5)	7(7)	52(9)	28(4)	50(6)	50(6)
Hoppenfeld, (1976)	-----	----	f & m	120	30	45-50	-----	45	35
Magee, (1987)	-----	----	f & m	110-120	10-15	30-50	30	40-60	30-40
Kendall & McCreary, (1983)	-----	----	f & m	125	10	45	10	45	45
Mundale, Hislop, Rabideau & Kottke, (1956)	20 - 30	20	f	-----	12.4	-----	-----	----	----
		16	m	-----	10.4	-----	-----	----	----
Roach & Miles, (1991)	25 - 39		f	123	22	44	---	36	33
			m	123	22	46	---	33	34
Simoneau, Hoenig, Lepley & Papanek, (1998)	18 - 27	39	f	----	----	----	----	46(13)	38(9)
	18 - 26	21	m	----	----	----	----	44(7)	32(9)

The resting position of the hip is defined as that position where the capsule and ligaments surrounding the joint are in their most slack position, thus offering the least resistance to movement. In the hip, this occurs at approximately 30° of both flexion and abduction, with slight external rotation (ER) (Gray, 1974). Mechanically, this is also referred to as the neutral zone (NZ): that part of the range of motion in which there is minimal resistance to motion (Panjabi, 2003). Conversely, the close-packed position, in which the NZ is reduced to its smallest value due to soft tissue tension, is in full extension, internal rotation, and abduction (Magee, 1987). The strongest ligament around the hip joint is the Y-shaped ilio-femoral ligament on the anterior aspect, which is intimately connected to the joint capsule. From its proximal attachment to the AIIS, this ligament divides into two strong bands (superior and inferior), which attach to the upper and lower regions of the intertrochanteric line, with the more lateral band also attaching to the neck of the femur (Figure 2-1).This ligament has been shown to withstand higher tensile stress with less material strain at failure than other hip ligaments (J. D. Hewitt, Glisson, Guilak, & Vail, 2002), and a lower strain value than the inferior gleno-humeral ligaments: 6.2% (J. Hewitt, Guilak, Glisson, & Vail, 2001) compared to 10.9% (Bigliani et al., 1992)or 9.3% (Ticker et al., 1996). Together with the pubo-femoral ligament, they resist extension of the femur past a neutral position, thus providing a strong passive support that allows erect posture to be maintained without active muscular hip control. The ilio-femoral ligament also plays a role in limiting hip lateral rotation, abduction, and adduction (the latter when combined with flexion).

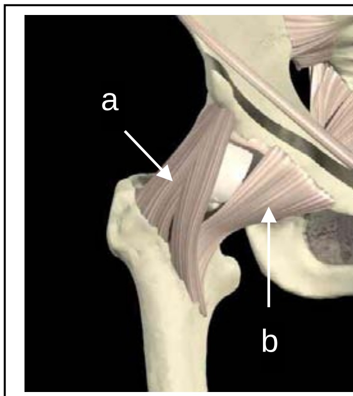


Figure 2-1: Anterior aspect of the hip joint, showing the a) ilio-femoral ligament and b) pubo-femoral ligament (courtesy of Primal Pictures Ltd).

Medial rotation is limited more so by the posterior capsule and ischeo-capsular ligament: a thickening in the capsular fibers that travel from the ischeum along the inferior hip joint capsule (Gray, 1974).

In addition to the ligaments, muscles also play a role in controlling the limits of hip motion. With the knee extended, hip flexion is usually limited by the hamstrings (biceps femoris (BF), semitendinosus (ST) and semimembranosus (SM)). Hip extension may be limited by either one or two-joint muscles. The one-joint muscles include psoas and iliacus. Psoas, however, is not a true one joint muscle in that it crosses numerous lumbar vertebrae, thus contralateral spinal side-bending is required to elicit a true psoas stretch (McGill, 2004). Sartorius, tensor fasciae latae (TFL) and rectus femoris (RF) are all considered two joint muscles, in that they also cross the knee (indirectly in the case of TFL), thus will be affected by knee position. Due to their individual alignments, sartorius will be in more tension when hip extension is combined with internal rotation, TFL with external rotation, and RF is more specifically tested when in a position of neutral hip rotation. The medial rotators of the hip include TFL, gluteus minimus (GMin) and the anterior fibers of gluteus medius (GMed); increased tension in any of these muscles may be responsible for limiting external rotation. External rotation is accomplished by piriformis, quadratus femoris, obturator internus and externus, gemellus superior and inferior, as well as the posterior fibers of gluteus medius: any of which may limit internal rotation (Kendall et al., 1983).

2.1.2. Sacro-iliac joint and pelvis

Forces transmitted in a cephalaud direction from the hip joint travel through the lateral portions, or ilia of the pelvis, also referred to as the innominate bones. These two large bones articulate posteriorly with the lateral edges of the sacrum at the sacro-iliac (SI) joint. Being shaped like an inverted pyramid, the sacrum sits within a wedge-shaped opening, formed between the two ilia. Most of the support for the joint is provided passively by its actual configuration, as well as the large number of strong ligaments. Muscular support of the SI joint is supplied in part by a few fibers of gluteus maximus (GM), which cross the SI joint (although most of the fibres of this muscle originate lateral and/or inferior to the joint, thus providing no direct support). Piriformis, originating on the anterior surface of the sacrum, and inserting on the head of the femur, provides the only other active muscular support to the SI joint (Mitchell, 1999).

Because the pelvis sits atop the two ball and socket joints of the hips, its position in space is highly dependent on associated soft tissue stiffness and strength. Increased tension in the hamstring (HS) muscles will pull on the posterior/inferior aspect of the pelvis, resulting in a more posteriorly tilted position, unless a corresponding increase in anterior hip muscle tension counteracts this effect. Conversely, too much tension in the hip flexors (iliacus, RF) may pull the anterior portion of the pelvis down, resulting in an increased anterior tilt. This moves the base of the lumbar spine more anterior to the hip joints, resulting in a more anterior position of the COM of the head, arms, and trunk. To compensate for this, an increased lumbar lordosis is required (Whittle & Levine, 1995) or increased activity in the hip and back extensor muscles.

Using X-ray analysis, a normal pelvic angle in upright standing is approximately 30 - 32°, calculated by drawing a line along the top of the sacrum in a sagittal orientation, bisecting the horizontal (Magee, 1987; McGill & Norman, 1986). However, in a clinical setting, seldom is a numerical value attributed to the actual pelvic or lumbo-sacral angle, due to the difficulty of obtaining reliable measurements. Instead, more subjective observations are noted, with regards to the relative position of the ASIS and PSIS landmarks, and apparent sagittal tilt of the pelvis. In a research setting, using external markers and a VICON motion analysis system, Schache et al (2003) found a significant gender difference in standing anterior pelvic tilt angle, with females averaging 8.7 (3.8)° compared to 5.4 (3.3)° for males, when a line drawn from the ASIS to the PSIS bisected the horizontal (Schache et al., 2003). Other researchers have shown this same angle to be similar between sexes, measuring an average of 12° and 11.8 (4.4)° (Smidt, McQuade, Wei, & Barakatt, 1995; Whittle et al., 1995). Exaggerated anterior or posterior tilting of the pelvis resulted in measurements of 23.0 (4.8)° and 3.2 (5.5)°, respectively, resulting in a total sagittal excursion of the pelvis of approximately 26° (Whittle et al., 1995).

2.1.3. Lumbar spine

The lumbar spine is comprised of 5 bony vertebrae, separated by fibrocartilaginous discs. The L5, S1 segment transfers forces and motion from the pelvis to the upper body (and vice versa) via a 3-point weight-bearing system: the intervertebral disc and the two facet, or zygo-apophyseal, joints. In that the facet joints lie posterior to the disc, they tend to transfer more of the upper body weight when the lumbar spine is in extension (Adams & Hutton, 1980). Conversely, small amounts of forward flexion will increase the intra-discal pressure and lessen the facet joint compression (Shirazi-Adl & Drouin, 1987; Shirazi-Adl, 1994). During a gait cycle, the total torque transmission through L4, L5 oscillates between the facets and disc: at heel strike, when the pelvis is maximally rotated axially, the facets engage and transmit virtually all of the torque. At mid-stance, when the facet joints have a larger neutral zone and little approximation, the discs become the main structure for force transmission (Gracovetsky, 1997).

Orientation of the facet joints will dictate the amount of movement in the 3 planes of motion. However, there is a large individual and segmental variability as to the alignment of the facet joints, which will ultimately affect associated motion (Taylor et al., 1986). Seminal works by Pearcy et al (1984; 1984) used X-ray analysis to identify the amount of individual vertebral motion between L1 and S1 in healthy individuals when upright standing (Table 2-2).

Table 2-2: Mean values of active spine motion (degrees) in the sagittal, frontal and horizontal planes, as determined by X-ray analysis. Sagittal values include the SD of the measurement, whereas the frontal and horizontal means are followed by the range of variability. (Pearcy & Tibrewal, 1984; Pearcy et al., 1984)

Level	Flex.	Ext.	Total sag.	Side bend R	Side bend L	Total frontal	Rot. R	Rot L	Total horiz.
L1, L2	8 (5)	5 (2)	13 (5)	-5(-8 to -2)	6(4 to 10)	11	-1(-2 to 1)	1(-1 to 2)	2
L2, L3	10 (2)	3 (2)	14 (2)	-5(-8 to-4)	6(2 to10)	11	-1(-2 to 1)	1(-1 to 2)	2
L3, L4	12 (1)	1 (1)	13 (2)	-5(-11 to 2)	5(-3 to 8)	10	-1(-3 to 0)	2(0 to 4)	3
L4, L5	13 (4)	2 (1)	16 (4)	-3(-5 to 1)	2(-3 to 6)	5	-1(-2 to 1)	2(0 to 3)	3
L5, S1	9 (6)	5 (4)	14 (5)	0(-2 to 3)	-2(-6 to 1)	2	-1(-2 to 0)	0(-2 to 2)	1
Total spine	41	16	70	-17	18	35	4(1 to 6)	5(1 to 11)	8

There are a few specifics worth noting with regard to these measurements. Although the total amount of sagittal motion in the lumbar spine at each segment is approximately 14°, there is minimal extension at the middle levels; most of the available movement is into flexion. This is likely due to an increase in the standing position lumbar lordosis at these levels, resulting in fewer residual degrees of extension available with motion (Table 2-3). Having said that, the L5, S1 vertebral level demonstrates a much larger angle of lordosis than any of the other levels, sitting in approximately 21 - 27° more anterior rotation than L4, L5 (Kingma, Bosch, Bruins, & van Dieen, 2004; Pearcy et al., 1984). The L5 vertebra is characterized by a thicker anterior body than posterior, adding to the lordotic curve (Gray,

1974) and has a smaller spinous process posteriorly, allowing the increased extension/lordosis to occur without abutment of the posterior processes. The L5, S1 segment also demonstrates the highest variability in sagittal plane motion, with no consistent ratio of flexion/extension at this joint being found. Sitting atop the pelvis, it is the first vertebral level to adjust to changes in the position of the pelvis: an 11° increase in anterior pelvic tilt can result in a corresponding 10° increase in lumbar lordosis. Conversely, a 7.6° loss of anterior pelvic tilt can diminish the lordosis by 10° (Pearcy & Whittle, 1982). With regards to other planar movements, most of the frontal plane motion occurs at the upper lumbar segments, with significantly less occurring at L4, L5 and L5, S1 (mean of 6° and 3°, respectively). Lumbar rotation, in the horizontal plane, is greatest between L2 and L4, but still less than 2° in each direction (Table 2-2) (Pearcy et al., 1984).

Table 2-3: Angle of lordosis at individual vertebral levels: averaged over 11 participants in upright standing: mean (SD) (Pearcy et al, 1984a)

Lordosis in degrees: mean (SD)	
L1, 2	2 (5)
L2, 3	7 (4)
L3, 4	11(3)
L4, 5	17 (5)
L5, S1	38 (6)
Total lumbar spine	74 (7)

Pure planar movements of the lumbar spine are rare, due to the non-planar orientation of the facet joints; the anteriomedial third of the facet joints are generally oriented coronally (in the frontal plane), and the posterior portion is more sagittal (J. R. Taylor & Twomey, 1986). This results in “coupled motions” of the spine: for example, axial rotation of the lumbar spine is accompanied by a certain amount of side bending, or frontal plane motion, which varies with the segmental level. In the upper lumbar spine, axial rotation is accompanied by ipsilateral side bending (i.e. right rotation goes with right side bending). However, at the L5, S1 segment, this pattern changes, such that rotation is now accompanied by contralateral side bending. The L4, L5 segment, seen as a transitional level, demonstrates no consistent pattern of coupling (Pearcy et al., 1984b). Given that this level has, on average, the greatest sagittal and least axial rotation (N. Bogduk & Twomey, 1991; Pearcy et al., 1984), and is positioned near the apex of the lumbar lordosis as a transitional segment, it is not surprising that L4, L5 has the highest incidence of wear and tear in the lumbar spine (Macnab, 1977).

In addition to guiding movement, the facet joints have been shown to limit the range of motion. As shown in Table 2-2, approximately 1- 2° of axial rotation occurs to each side, at each lumbar vertebral segment. The facet joint on the side of compression (e.g. the right facet joint with left axial rotation of the superior segment) typically is the first structure to limit rotation in cadaveric specimens (Adams & Hutton, 1981). The amount of rotation has been shown to increase proportionately with increasingly degenerated discs, which the authors suggested might have more to do with thinning of the facet joint cartilage (thus more space between the bony surfaces) than actual loss of disc height. Once abutment of the facet joint occurs, the vertical axis of rotation changes, from one passing through the posterior aspect of the disc to an axis at/through the compressed facet joint. As a consequence, the contralateral facet joint now experiences greater tensile stresses to the capsule and ligaments. Repetitive or excessive compression (associated with rotation greater than 3°) to a facet joint can result in failure, likely a collapse of the articular cartilage or the subchondral bone. Of all three planar movements, axial rotation has been shown to generate the largest contact forces in the facet joints (Shirazi-Adl, 1991), when compared with identical amounts of sagittal or coronal motion; this is not surprising considering the small amount of rotation available (Table 2-2), and is indicative of the orientation and guiding role the facets play in lumbar motion. According to Bogduk and Twomey (1991), 65% of the resistance to axial torsion is provided by the posterior elements of the neural arch: the impacted facet joint providing most of this resistance, followed by the contralateral facet joint under tension and the supraspinous and interspinous ligaments. Only 35% of the resistance to torsion is provided by the intervertebral disc (Bogduk et al., 1991).

Using cadaveric specimens to analyze tissue damage associated with lumbar extension, Adams et al (1988) found that the first structure to be affected is the interspinous ligament, as it is squeezed between the spinous processes. However, if those processes are widely spaced, the facet joints will instead be heavily loaded with increasing extension (Adams et al., 1988). Pressure within these joints increases significantly with narrowing of disc space and increasing extension (Dunlop, Adams, & Hutton, 1984). In slight lumbar extension, as found in normal upright standing, the facet joint resists about 16% of the axial compressive force between vertebrae, with the 3 lower lumbar facet joints carrying higher loads than the upper ones, possibly due to the increasing lordotic angle in the lower spine (Table 2-3) (Adams & Hutton, 1980). Most of this compression and transfer of force occurs in the anteriomedial portions of the facets joints, due to the bony architecture of the joint.

Perhaps surprisingly, it appears that the facet joints may also play a role in limiting forward flexion. Although the erector spinae muscles actively control eccentric flexion against gravity, they may not always be active, as in the case of flexion/relaxation phenomena (Dickey, McNorton, & Potvin, 2003; Gupta, 2001; Olson, Solomonow, & Li, 2006; Solomonow, Baratta, Banks, Freudenberger, & Zhou, 2003), or when flexion is occurring in a side-lying or supine position. Tension will then be taken up by the strong capsular ligaments, the ligamentum flava and supra and infraspinous ligaments (Adams, Hutton, & Stott, 1980). However, along with pure flexion around a coronal axis, spinal flexion also involves a 1-3mm forward translation of the uppermost vertebrae on the one below it, with the least amount of translation occurring at the L5, S1 segment (N. L. T. Bogduk, 1997; Kanayama, Abumi, Kaneda, Tadano, & Ukai, 1996). The anteriomedial facet joints, in this case, are most important to guide the motion, and resist extreme forward glide, or shear (Adams et al., 1988; Shirazi-Adl, 1991; Twomey, 1983) which is known to be linked with increased injury risk (Norman et al., 1998).

Other authors have measured in-vitro lumbar spine motion in a healthy population by means of some form of imaging or an external measurement system. Taylor & Twomey (1980) found that the average amount of spinal flexion in cadaver specimens was greater than that measured in-vivo, and suggested that the passive properties of erector spinae, as well as intra-abdominal pressure, may be responsible for this difference. In the majority of studies, spinal motion is measured between the T12 and S1 segments. As can be seen in **Error! Reference source not found.**, in addition to large inter-subject variability, there is also great variability between studies, which can be partly explained by the different positions/methods utilized. While Sullivan et al (1994)u{{691 Sullivan,M.S. 1994/a;}} used a kneeling fully flexed position to measure flexion, and a prone/extended position for extension, all other authors used measurements taken in standing. Clayson et al.(1962) was the only group to use X-ray imaging to quantify motion; Taylor and Twomey (1980) used a spondylometer, McGregor et al. (1995) used a triaxial potentiometer, and the others used inclinometers. Even so, the total amount of sagittal spine motion was fairly similar. McGregor et al. (1995) and Sullivan et al. (1994) both found a significant gender effect for mean flexion and extension angles, as well as declining amount of motion with age (not included in the table).

Table 2-4: Normal range of average lumbar motion (SD) in degrees, as documented by various authors.

Author	sex	age	flexion	extension	total sag.	side bend	rotation
McGregor, McCarthy & Hughes, (1995)	M	20 - 29	64.9(9.7)	30.2(8.0)	94.9	35.5(5.5)	28.6(8.2)
	F		54.5(9.3)	26.1(6.8)	80.6	33.7(6.4)	30.5(8.3)
Sullivan, Dickinson, & Troup, (1994)	M	16 - 34	32(9)	53(10)	85	-----------	-----------
	F		25(8)	61(10)	86		
Clayson et al, (1962)	F	approx 18 - 25	-----------	------------	92	----------	-----------
Taylor & Twomey, (1980)	M	20 - 35	42(6.3)	-----------	-----------	-----------	33(6.2)
	F		42(6.7)				33(6.0)
Tsai & Wredmark, (1993)	F	25 - 43	58 (14)	17(7)	75(16)	-----------	-----------

In addition to analyzing the bony architecture of the spinal column, the intervertebral discs must also be considered for the role they play in force transfer. This is highly variable with the posture, as was demonstrated in early works by Nachemson and Morris (1966; 1964). Using a pressure transducer in-vivo to measure intra-discal pressures in various positions, the authors found that the pressures in standing and reclining were, on average, 30% and 50% lower, respectively, than those found in sitting. As previously mentioned, in slight lumbar extension (i.e. upright standing), 16% of the compressive load is carried by the posterior facet joints, whereas in slight flexion, the load is carried almost exclusively by the discs (Adams et al., 1980).

2.2. Dynamic applied anatomy: gait

The human body, however, is not a series of joints and segments working independently. The body is a dynamic system, with contractile, elastic, and inertial properties, which allow us to move, initiate and respond to perturbations, and adapt to changes in the external environment. For the purpose of this research, kinetics and kinematics associated with gait will be used as the template on which to compare other activities. Consequently, a thorough understanding of the biomechanics of gait is essential.

2.2.1. Variability of gait

Many authors have commented on the large inter-subject variability associated with gait patterns (Crosbie, Vachalathiti, & Smith, 1997b; Rowe & White, 1996), yet intra-subject variability is generally low (Growney, Meglan, Johnson, Cahalan, & An, 1997; Schache et al., 2002). This is obvious to the curious observer, as most of us can be identified from a distance by the coordination patterns that dominate our walking style. Thus, despite large variability between people, we each adopt a walking style that uniquely optimizes efficiency and comfort.

2.2.2. Kinematics of the hip

Using upright standing as the zero reference, normal gait at free walking speed requires approximately 40°- 50°of total flexion/extension at the hip (Benedetti, Catani, Leardini, Pignotti, & Giannini, 1998; Crosbie, Vachalathiti, & Smith, 1997a; Johnston & Smidt, 1969; Kerrigan et al., 2001; Murray, 1967; Perry, 1992; Torry, Schenker, Martin, Hogoboom, & Philippon, 2006). While Kerrigan, et al (2001) describe the relationship of flexion/extension as 24°/20°, most other authors refer to the balance as more of a 30°/10° split. Assumedly, these differences are due to the definition of the zero reference; a more anterior tilted pelvis would result in a greater relative degree of flexion at the hip. Peak hip extension occurs at approximately the same time as contralateral heel strike, whereas flexion peaks during the ipsilateral mid-to-late swing phase (Perry, 1992).

In the coronal plane, ranges of motion are small, and the literature is varied. According to Perry (1992), the hip is in approximately 10° of adduction at heel strike, which changes to 5° of abduction during early swing. Other authors have documented ranges of 4 - 5° of adduction, and 6 - 7° abduction (Benedetti et al., 1998; Johnston et al., 1969). Obviously, these measurements are of a small magnitude, and therefore a difference of a few degrees of error in measurement will have a larger impact.

Axial rotation at the hip joint involves a total arc of approximately 8°, with internal rotation peaking at contralateral toe-off, and external rotation at ipsilateral toe-off (Benedetti et al., 1998; Johnston et al., 1969; Perry, 1992). However, these motions are not measured relative to the pelvis motion; when pelvis rotation is also added to the joint, the total hip joint rotation occurring within the acetabulum increases to 15° (Perry, 1992).

2.2.3. Kinematics of the pelvis

During each stride, the pelvis moves within a range of 2 - 6° in the sagittal plane, with a mean of approximately 4°, oscillating equally in each direction around a mid-point of approximately 10 - 12° relative to the horizontal (method previously described) (Benedetti et al., 1998; Crosbie, Vachalathiti, & Smith, 1997a; Perry, 1992; Smidt et al., 1995; Thurston & Harris, 1983; Thurston, 1985; Whittle & Levine, 1995). Maximum anterior pelvic tilt occurs at roughly the same time as toe-off, and corresponds temporally with acceleration of the swing leg, and maximum lumbar lordosis (Thurston et al., 1983). The amount of anterior pelvic tilt has also been shown to significantly correlate with degrees of hip extension at a comfortable walking speed (Kerrigan et al., 2001; L. W. Lee et al., 1997). Conversely, maximum posterior pelvic tilt corresponds with lumbar flexion and lower thoracic extension at heel-strike (Crosbie, Vachalathiti, & Smith, 1997b). The mid-point around which the sagittal motion occurs varies greatly between people (Rowe & White, 1996), with total sagittal motion decreasing with age (Kerrigan et al., 2001).

Frontal plane motion of the pelvis takes the form of an ipsilateral dip of the pelvis on the swing side, corresponding with, but slightly lagging behind, toe-off. Average pelvic motion has been described as being between 4.0 and 7.0° per side (Crosbie et al., 1997a; Perry, 1992; Thurston, 1985; Thurston et al., 1983). This movement is controlled by the hip abductor muscles on the stance side; lack of muscle strength may result in the typical "trendelenburg sign" whereby the pelvis on the swing side drops down during single leg stance, requiring compensatory lumbar lateral bending to avoid an inefficient side to side (coronal) motion of the head and trunk (Booher & Thibodeau, 2000; Magee, 1987). Conversely, abductor weakness may also present as a shift of the upper body centre of mass (COM) over the stance leg, thus decreasing the adductor moment at the hip, lessening the required abductor activity. Either way, frontal motion of the pelvis corresponds to a contralateral side bending of the lumbar spine, although the latter has been found to generally lag behind the pelvis slightly with regards to timing (Thurston & Harris, 1983).

Although Perry (1992) describes the largest arc of pelvis motion during gait as occurring around a vertical axis, there appears to be great variability between subjects and also amongst published literature. As the pelvis rotates horizontally with each step, it helps to increase the stride length prior to heel strike. Peak rotation corresponds with heel strike, with ranges of 1.6 – 10.1° in each direction being found in the literature (Benedetti et al., 1998; Crosbie et al., 1997a; Perry, 1992; Rowe et al., 1996; Thurston, 1985; Thurston et al., 1983). Numerous authors have, in fact, noted that axial rotation data have greater variability than sagittal or coronal motion (Benedetti et al., 1998; Crosbie et al., 1997a; Crosbie et al., 1997b; Murray, 1967). This may be due in part to the fact that large deviations from constrained motion in the other two axis would result in energy inefficiencies: i.e. too much flexion/extension or side bending would displace the upper body COM, requiring greater muscle energy expenditure to maintain whole body stability. Large amounts of rotation, on the other hand, would tend to lengthen the stride, but not necessarily displace the upper body COM. In addition, three dimensional modeling of the hip, back and pelvis generally follows a Cardan sequence of flexion/extension, ab/adduction, followed by axial rotation. In that the errors associated with each of these motions tend to accumulate as the computations progress, pelvis rotation may demonstrate greater variability simply because of the increased error of the third axis associated with mathematical modeling.

Smidt et al (1995) measured the pelvis and sacro-iliac motion in 32 young adults (15 men, 17 women) during a straddle position, similar to a sagittal “splits”. Measurements were taken at 90% of their maximum available range. In this extreme position, the pelvis rotated anteriorly (sagittally) 11° and exhibited 17° of axial rotation (relative to upright standing) toward the side with the lagging leg. Coronal motion was the smallest, rotating an average of 6° to the left, and 3° to the right, with right and left straddle positions, respectively. In a situation of high-speed running, where near maximum straddle positions might be achieved, this pelvic motion would have to be accommodated by the thoraco-lumbar spine, in order to keep the head facing forwards. That is, one could expect 11° of thoraco-lumbar extension, 17° of contralateral axial rotation, and somewhere between 6° and 3° of contralateral side bending. The mean between-hip angle (measured from one femur relative to the other) in the 90% straddle position was 93(27)° (Smidt et al., 1995).

2.2.4. Kinematics of the spine

The lower spine tends to respond to movements of the pelvis, both temporally and quantitatively (Crosbie et al., 1997b). Sagittally, the lumbar lordosis changes only approximately 2 - 3° in each direction during normal walking, for a total range of 3 - 6° (Callaghan, Patla, & McGill, 1999; Crosbie, Vachalathiti, & Smith, 1997a; Thurston, 1985;

Whittle & Levine, 1995). This corresponds closely with the 2 - 6° of pelvic anterior/posterior tilt previously mentioned. Maximum lordosis tends to occur at the same time as peak anterior pelvic tilt of the pelvis, occurring with toe-off, although the movement of the pelvis precedes the lordosis slightly (Thurston & Harris, 1983) indicating that the lumbar spine motion may be driven by the pelvis.

Coronal spine motion during gait compensates for the contralateral side bending of the pelvis, thus allowing the body's centre of gravity to stay within the base of support. Thurston and Harris (1985) measured spinal side bending as ranging between 3 - 14° to each side, with mean angles between 3.1° and 8.5° being documented by other authors, again indicating the large variability found in spine motion (Callaghan et al., 1999; Crosbie et al., 1997a; Rowe et al., 1996; Thurston, 1985; Thurston et al., 1983).

Axial twist of the lumbar spine during walking tends to be similar to or slightly less than that of the pelvis: 4.0 – 8.3° (Callaghan et al., 1999; Crosbie et al., 1997a; Rowe et al., 1996; Thurston et al., 1983). Assumedly, the remainder of the motion required to keep the head and eyes facing forwards will be supplied by the thoracic and cervical spine, both of which have facet joint alignment that allows more axial rotation (Gray, 1974). Variability has been found to be greater in axial twist compared to the other two planar movements (Crosbie et al., 1997b; Rowe et al., 1996), with many people appearing to rotate more in their lumbar spines than their hips, possibly due to limited hip mobility or specific motor patterns they have adopted. For those people who rotate 8.3° to each side, approximately half the available lumbar rotation of 14 or 15° (McGregor et al., 1995) (**Error! Reference source not found.**) is already being utilized for a regular walking pace. Axial rotation has been shown to occur slightly before that of the pelvis (Thurston et al., 1983), such that the upper body seems to be driving the pelvis, possibly due to the influence of arm swinging.

Little has been said so far about the thorax and its relation to the hip/pelvis/lumbar spine. At normal walking speeds, there is a strong coupling between the thorax and pelvis. That is, right rotation of the thorax, associated with the right swing leg, corresponds with a right rotation of the pelvis, thus very little lumbar twist. However, at higher velocities, although the thorax and thigh remain in an anti-phase pattern, the pelvis motion becomes more synchronous with the thigh. Now right thorax rotation and right thigh swing through are accompanied by a left rotation of the pelvis, resulting in more axial twist between the thorax and pelvis (i.e. in the lumbar spine) (Bruijn, Lamoth, Kingma, Meijer, & van Dieen, 2006). This change in pelvis alignment with regards to the thigh would also assist in lengthening the stride, a component of increasing velocity. Although more variable than the rotations at the hip and knee, the average amount of thorax rotation to each side at self-selected speed has been documented as 2.5 - 6.8° and 8.9° (Crosbie et al., 1997a; Murray, 1967).

2.2.5. Kinetics of the spine

Although kinematics provide us with the information about body movement, and confirms numerically what we see visually, it is the information that we cannot see that may be more important in determining injury risk and trauma: the forces associated with the spine. Since in-vivo measurement is not a reasonable option, we turn to mathematical modeling to assist us in understanding and predicting the loads in the spine.

Rigid link segment models, using ground reaction forces and kinematic data, allow us to calculate the moments and forces at each of the lower extremity joints and up into the spine, with most models using the L4, L5 vertebral segment as the focus for these kinetic outcomes. Although this same modeling technique can take place from the top down, the large size of the non-rigid trunk, with its varying organ densities and COM position, result in outcomes with more variability, thus favoring the bottom up approach when possible (Kingma, de Looze, Toussaint, Klijnsma, & Bruijnen, 1996). Using a four muscle equivalent model and a top-down approach, Capozzo (1984) found maximum L3, L4 compression forces averaging 1.6 times body weight during regular walking (Cappozzo, 1984).The addition of EMG data from relevant trunk muscles allows calculation of the bone-on-bone forces, which have been found to be three times greater than the joint reaction forces determined by rigid link modeling, averaging in the range of 2.18, 0.12 and 0.8 times body weight (BW) for maximum compression, A/P shear and lateral shear, respectively, during walking (Callaghan et al., 1999). Increasing their speed by 20 steps/minutes caused an increase in each of these variables, to 2.45, 0.16 and 0.10 X BW, respectively, which would be approximately 1682 N, 110 N and 69 N for a 70 kg person. This compression amount is well below the NIOSH action limit of 3400N (Dept of Health and Human Services (NIOSH), 1981), as well as that found in upright sitting (2128 N)(Kavcic, Grenier, & McGill, 2004a).

2.3. The effect of aging

The only hip motion that has been shown to significantly decrease with age (at least up to the age of 74) in a healthy population is hip extension, both in passive measurements and during gait (Kerrigan et al., 2001; Roach et al., 1991). Stride length, however, and thus total sagittal motion when walking, has also been shown to decrease, with most of the loss occurring in the direction of extension (Benedetti et al., 1998; Karamanidis & Arampatzis, 2007; Kerrigan, Todd, Della Croce, Lipsitz, & Collins, 1998; Murray, 1967). This is accompanied by a decrease in self-selected walking speed, which may be due more to the shortening of the step than an actual change in cadence (Burnfield, Josephson, Powers, & Rubenstein, 2000; Crosbie, Vachalathiti, & Smith, 1997a; Riley, Della Croce, & Kerrigan, 2001).

The posterior-lateral aspect of the lumbar facet joints show the most degenerative changes with age (J. R. Taylor & Twomey, 1986). In stationary measurements, total lumbar motion in all 3 planes decreases with age. Using a tri-axial potentiometer during active ROM, McGregor et al (1995) found mean decreases of 36.7°, 19.0° and 9.1° in total sagittal, coronal and axial active ROM respectively, when comparing men in their twenties to those in a 60 – 70 year age bracket. During gait, Thurston & Harris (1983) found sagittal displacements in both the spine and pelvis significantly decreased with age, as did axial rotation of the spine. However, normal gait does not require the full range of lumbar motion. Although smaller amounts of lumbar motion may appear to be associated with aging, it has been suggested that the change may be due more so to the decrease in stride length or speed, and that if this were standardized, the actual kinematic measurements in the spine and pelvis might be very similar across the age groups (Crosbie et al., 1997a).

2.4. The effect of gender

There is conflicting evidence as to whether a gender difference exists regarding hip motion. Roach and Miles (1991) found no significant differences, but a slight trend towards increased external rotation in females (Table 2-1). Simoneau et al. (1998) describe significantly more hip external rotation in the female population, whereas Gombatto et al (2006) and Mellin (1990) found the trend to be opposite: slightly greater in males. Larger amounts of active hip flexion with a straight knee (thus longer/more flexible hamstrings) have been documented in females (Bach, Green, Jensen, & Savinar, 1985; Dolan & Adams, 1993) as well as more extension, abduction and internal rotation in the female population (Bach et al., 1985; Dolan & Adams, 1993). However, in the Bach et al. study, this was only present on one leg, adding to the confusion of generalized statements about gender differences with regards to hip joint mobility.

Controversy also exists regarding differences in the amount of available spinal motion between the sexes. Taylor and Twomey (1980) found that adolescent and young females have more sagittal and horizontal spine motion than males of the same ages, but these differences disappear in the middle age and elderly groups. Conversely, McGregor et al., (1995) found spinal motion in all planes except right side bending to be significantly less in females. Sullivan et al., (1994) using sitting flexion and prone extension, found that males demonstrated more flexion, females more extension, yet when the total sagittal motion was calculated, females exhibited only slightly more range than males (Table 2-4).

When hip and back are looked at as a grouping, active hip external rotation in prone lying precipitates associated lumbar rotation earlier in men than in women. The authors suggest that increased muscle mass and overall passive stiffness in the male hip joints may be responsible for this difference in timing, as the total range of hip mobility was not significantly different (Gombatto et al., 2006).

During walking, however, a few significant differences between the genders have been noted. At self-selected speeds, females tend to walk with a slightly faster cadence and shorter step length than males (Crosbie et al., 1997a; Schache et al., 2003), using significantly more hip flexion (Kerrigan, Todd, & Della Croce, 1998; Schache et al., 2003). When normalized for subject height, though, female step length was equal to or greater than that demonstrated by the men (Kerrigan, Todd, & Della Croce, 1998; Schache et al., 2003; Williams K.R., Cavanagh, & Ziff, 1987). With this in mind, gender differences in gait kinematics must be taken with a “grain of salt”, knowing that cadence and step length differ, and may be due to an average difference in leg length between males and females. Nevertheless, females tend to walk and run with a slightly more anteriorly tilted pelvis, and rotate their hips and pelvis axially more than males (Crosbie, Vachalathiti, & Smith, 1997a; Ferber, McClay Davis, & Williams, 2003; Hurd, Chmielewski, Axe, Davis, & Snyder-Mackler, 2004; Pollard, Sigward, & Powers, 2007; Schache et al., 2003). In running, women tend to have slightly more hip flexion than the men, and significantly more hip adduction (Ferber et al., 2003; Schache et al., 2003), which may be due partly to a shorter average femur length.

In the lumbar spine, males have been shown to side bend more than the females when walking, whereas females consistently demonstrate more thoracic axial rotation. This gender difference becomes greater with increasing velocity, as females exhibit even more axial rotation (Crosbie et al., 1997a). When the velocity increases to a running speed, females have more movement in all 3 planes than the male counterparts (Schache et al., 2003).

2.5. The effect of velocity

Increased stride length, associated with faster walking speeds, is mainly accomplished by an increase in the amount of hip flexion (Murray, 1967). This is accompanied by a small increase in overall anterior pelvic tilt during the cycle (Murray, 1967) , which is then associated with an increase in the lumbar lordosis and/or a forward lean of the trunk (Callaghan et al., 1999; Crosbie et al., 1997a; Murray, 1967), as well as lateral bending (Crosbie et al., 1997a). Interestingly, there seems to be a carry-over effect in those who habitually use higher velocities: when comparing runners to non-runners walking at identical speeds, the runners walked with a greater stride length and swing duration, with a decreased stride frequency (Karamanidis et al., 2007).

When increased to a running speed, hip motion in all 3 planes increases compared to walking speed. Sagittal motion when running has been documented as 71 and 74° for males and females, respectively, thus approximately 20 – 30° greater than in walking. Likewise, coronal and axial rotations for the hip range from 23 -29° and 33 - 37°, respectively (compared to 11° and 15° when walking) (Benedetti et al., 1998; Johnston et al., 1969; Perry, 1992; Schache et al., 2003).

Pelvic motion in all 3 planes increases significantly with faster walking speed, although the differences between self-selected and fast walking were slightly less for axial rotation and ranged from 5.7 – 9.7° (Crosbie, Vachalathiti, & Smith, 1997a; Kubo, Holt, Saltzman, & Wagenaar, 2006). Similarly, changing from fast walking to a running speed increases the amount of all 3 planar motions even more so (Schache, Blanch, Rath, Wrigley, & Bennell, 2005).

In the lumbar spine, increased gait velocity results in significantly more lateral bend and flexion/extension, with a smaller trend for an increase in axial rotation (Callaghan et al., 1999; Crosbie et al., 1997a; Schache et al., 2003). However, when running, total axial rotation has been documented at 24 - 28° (male – female) (Schache et al., 2003), which is very close to the maximum available, as described by McGregor et al (1995) of 28.6° and 30.5° for males and females, respectively (Table 2-4).

Increasing pelvic rotation with faster speeds may result in a greater excursion of thoracic axial twist. In a study involving 60 healthy men, Murray found that axial rotation increased from an average of 6.8° to one of 8.9° when changing from moderate to fast walking speed (Murray, 1967). However, Kubo et al (2006) found a slight decrease in true thoracic rotation with increasing gait velocity, but the overall relative angle between the pelvis and thorax increased between 9.7° and 14.3°, mostly due to increasing pelvic rotation. Either way, the angle of rotation between the thorax and pelvis increases, and the lumbar spine bears the brunt of this motion.

2.6. Arm swing

During each stride, shoulder flexion/extension occurs over an arc of approximately 30 - 40° in time with contralateral anterior pelvis rotation. Shoulder extension is actively initiated by the posterior deltoid and teres minor, but flexion is mostly a passive activity at normal walking speeds (Murray, 1967; Perry, 1992). There is large inter-subject variability as to the extent of arm swing, as well as moderate discrepancy between right and left within the same person (Murray, 1967). Increasing velocity of gait results in a larger arc of arm swing in the sagittal plane, possibly to compensate for the larger stride length.

The largest effect of arm swing occurs around the vertical axis; the angular momentum of the arms during gait is opposite to that occurring in the rest of the body. When added together, the angular momentum around the vertical axis for the entire body remains close to zero (Cappozzo, 1983; Elftman, 1939). Thus, a larger stride length, with increased pelvis rotation and hip flexion, is counterbalanced with an increase in arm swing amplitude (Elftman, 1939).

Constraining arm motion during gait has the effect of lessening spine motion in all 3 planes, with the largest difference being seen in axial twist (Callaghan et al., 1999). It also results in increased activation levels of most trunk muscles except rectus abdominis (which, because of its vertical orientation has mainly a sagittal effect). Likely, this EMG increase is necessary to counter-balance the axial torque initiated with leg/pelvis swing. Subsequently, joint forces at the L4, 5 segment also tend to be higher with arm constraint (Table 2-5). Walking speed and stride length also decrease with unilateral or bilateral arm constraint (Eke-Okoro, Gregoric, & Larsson, 1997).

Table 2-5: Minimum and maximum joint forces at the L4,5 segment, calculated from an EMG driven model, averaged over 5 participants at normal walking speeds. Forces are normalized to body weight (joint force/body weight X 100) (Callaghan, Patla, & McGill, 1999).

	Compression				**A/P shear**				**Lateral shear**			
	arms		No arms		arms		No arms		arms		No arms	
	min	max	min	max	min	max	min	max	min	max	min	max
mean	120	218	125	233	-12	12	-15	13	-7	8	-9	10
SD	22	46	23	57	4	3	3	3	2	3	4	4

2.7. Muscle activation

Much of the literature examining muscle activation patterns during gait have focused on the lower extremity. Understandably so, in that activation levels as high as 85 and 75% of MVC are produced in the soleus and gastrocnemius, respectively, known to be important for both support and forward progression during gait (Perry, 1992; Zajac, Neptune, & Kautz, 2003). In the trunk, however, muscle activity remains relatively low during walking: the highest being in the range of 10(3)% and 11(3)% of MVC in the internal oblique and multifidus, respectively (Callaghan et al., 1999). As a trend, trunk muscle activation increases or stays the same with faster velocity, and will decrease with a slower cadence. As previously mentioned, trunk muscle activation levels increase when the arms are constrained (Callaghan et al., 1999).

2.8. Kinematic effects of limited hip mobility

Given the previous information about the importance of hip mobility in normal gait kinematics, it is important to understand what the implications might be when hip mobility is restricted. The capsular pattern of the hip, or the movements that decrease with capsular/ligamentous tightening, is a loss of flexion, abduction and internal rotation, with decreased internal rotation generally being the first limitation to occur with arthritis of the hip (Cyriax, 1975; Magee, 1987).

Habitual postures may play a role in determining which motions become diminished. In that mobility of a normal hip is not usually limited by bony approximation, muscles and non-contractile structures are the main limiting structures. Hip extension is often found to be less in the elderly, and may be due to a decrease in the frequency with which they obtain full extension more so than any active soft tissue tightening or bony blockage (Kerrigan et al., 2001). Prolonged sitting and decreased velocity of gait (thus decreased stride length and hip extension) may result in a gradual tightening of the anterior hip structures and eventual hip flexion contracture.

What are the effects of a stiff hip on structures higher up the kinetic chain? A significant increase in the sagittal and a decrease in coronal plane movements of the pelvis have been associated with osteoarthrosis of the hip (Murray, 1967; Thurston, 1985). Specifically, depending on the severity of hip flexor tightness, the pelvis may be pulled into an anterior pelvic tilt position during the latter or all of the stance phase (Perry, 1992). In a group of participants with neurologically-based hip flexion contractures, the degree of contracture correlated significantly with anterior pelvic tilt (L. W. Lee et al., 1997), which would, in turn result in a greater anterior tilt to the top of the sacrum: the base upon which the L5 vertebrae sits. Dunk et al (2009) has shown that lumbo-sacral flexion is driven by rotation of the pelvis in a “bottom-up” manner. If we assume the same phenomenon exists with pelvis extension, then it follows that the lower vertebral levels would be affected first, subsequently cascading up the spine. In that the lower 3 lumbar levels already carry more of the axial load, the lower facet joints would be more prone to over-loading than upper lumbar segments (Adams & Hutton, 1980). The corollary to this is that exercise programs aimed at increasing hip extension in the elderly have resulted in significant increases in hip extension both when measured statically and during comfortable walking speed, as well as a non-significant decrease in anterior pelvic tilt (Kerrigan et al., 2003). Similarly, a four week lower body stretching program in a group of long distance runners resulted in a decrease in total sagittal motion of the pelvis (Martin, Sommer, Ackermann-Liebrich, & Baumann, 1997).

It is, however, difficult to separate the lack of hip extension from corresponding changes that may have also occurred in pelvic alignment and the lumbar spine. People with osteoarthrosis of the hip joint may also have degenerative changes in the spine, limiting the compensatory mechanisms that the spine would normally make to offset effects of the stiff hip. Alternatively, prolonged sitting and slow walking, if responsible for a lack of hip extension, may also cause an increase in lumbar flexion, since a normal range of gait-related lumbar extension is not being obtained (Callaghan et al., 1999; Dunk & Callaghan, 2005). In a study of sheep-shearers, who spend up to 80% of their day in extreme hip and spine flexion, Milosavljevic et al (2005) found a 19% increase in hip flexion compared to a group of age-matched non-shearers, as well as a slight loss (1.4°) of hip extension and a 3.6°decrease in the lower lumbar lordosis (measured between L3 and PSIS). Thus, in this unique group, the prolonged posture caused changes in both joints; the change in lumbar mobility would not counter-balance the loss of hip extension, and would further flex the person into a forward position (Milosavljevic et al., 2005).

It therefore seems reasonable to seek out participants who have developed soft tissue hip tightness due to prolonged activities in flexion, but who are young enough that the changes may still be reversible. Anecdotally, there are many university-aged patients who appear in rehabilitation clinics with remarkably stiff hips, especially lacking extension and rotation (internal and external). Arthritic changes in the hip and spine would be rare in this age group, lending credence to the assumption that the restriction is due to soft tissue tightness, and thus amenable to stretching.

2.9. Hip motion and low back pain

Although a low back pain group is not part of this study, it is worthwhile mentioning how hip mobility may be a contributing factor to back pain. Most studies analyzing the interaction of low back pain and hip mobility utilize participants who are recruited because they have low back pain; the existing literature which asks people with stiff hips if they also experience back pain is largely limited to those with a neurological or aging basis to the hip stiffness, both of which may result in altered neuromuscular control of the spine.

In a low back pain population, a significant decrease in hip rotations (internal, external and total) has been demonstrated by Chesworth et al. (Chesworth, Padfield, Helewa, & Stitt, 1994) and Mellin (1988; 1990) (but in males only), yet other authors found no such difference (Bach et al., 1985; Paquet, Malouin, & Richards, 1994; Shum, Crosbie, & Lee, 2005). People with chronic low back pain have been found to exhibit less passive hip extension and active hip flexion than pain-free groups (Dolan & Adams, 1993; Mellin, 1988; Van Dillen, McDonnell, Fleming, & Sahrmann, 2000), perhaps due to long term slowing of gait and shortening of stride. In those with hip dysfunctions, 25 patients awaiting total hip replacement surgery all admitted to moderate or greater low back pain, which decreased significantly 3 months following surgery, and was maintained at the 2 year mark. Other than basic in-patient and home-based physical therapy post-operatively, they had received no specific therapy for their back (Ben-Galim et al., 2007). Hip flexor tightness has shown to be one of the few predictors of low back pain in adolescent boys (Kujala, Taimela, Salminen, & Oksanen, 1994).Similarly, reports of low back pain in adolescents have been shown to correlate with reduced hip flexion and lumbar extension (Kujala, Salminen, Taimela, Oksanen, & Jaakkola, 1992; Kujala, Taimela, Oksanen, & Salminen, 1997), decreased hamstring flexibility (Sjolie, 2004), as well as limited hip rotation (Fairbank et al., 1984). In addition to absolute movement, the relationship of spine to hip motion may be altered when low back pain is present: greater lumbar spine/hip motion ratio during the first 30° of forward flexion or extension has been documented (Esola, McClure, Fitzgerald, & Siegler, 1996; McClure, Esola, Schreier, & Siegler, 1997).Yet with coronal or axial motions, or functional movements such as putting on socks, reverse trends have been documented, with motion occurring at the hip sooner or to a greater extent, than the spine, in the low back pain group (Shum et al., 2005; Shum, Crosbie, & Lee, 2007; Wong & Lee, 2004).

Vad et al (2004) found that 33% of professional golfers surveyed admitted to a history of low back pain that limited performance for greater than 2 weeks within the past year. Of these, there was a significant correlation between a history of low back pain and decreased hip internal/external rotation on the lead leg, as well as decreased lumbar extension (Vad et al., 2004). These correlations were also demonstrated in a group of professional tennis players (Vad, Gebeh, Dines, Altchek, & Norris, 2003). Case studies exist that describe how corrective exercises aimed at normalizing hip mobility, while also decreasing shoulder turn in the golf situation, resulted in a reduction of low back pain within a 3 month period (Cibulka, 1999; Grimshaw & Burden, 2000).

Limited hip extension, resulting in increased anterior pelvic tilt, may ultimately cause increased lower lumbar lordosis, possible anterior shear of the posterior facets, thus lessening the foramenal space in the lower lumbar segments (Offierski & Macnab, 1983). This increased lumbar lordosis and spinal nerve root compromise have been associated with an increase in lower extremity soft tissue injuries in runners (Schache et al., 2005) and hamstring strains in footballers(Orchard, Farhart, & Leopold, 2004). Reports of low back pain in young female gymnasts correlated significantly with an increase in lumbar lordosis (mean of 41° compared to 35° in the no-pain group) (Öhlen, Wredmark, & Spangford, 1989). In the same study, an increased standing lumbar lordosis correlated negatively with the range of further lumbar extension available ($r = -0.69$), as well as the total range of sagittal motion ($r = -0.38$). Repetitive movements into this end range lumbar extension is believed to be causative of bony injuries such as spondylolysis and spondylolysthesis (Hardcastle et al., 1992; S. McGill, 2002; Schulitz & Niethard, 1980). Although a pure increase in sagittal lumbar mobility in adolescent girls was not predictive of low back pain, a high lumbar mobility / lumbar strength ratio has been shown to correlate significantly with reports of low back pain (Sjölie & Ljunggren, 2001).

2.10. Clinical measurement

The most frequently used method for measuring the lower extremity clinically, involves the use of a 360° goniometer (Figure 2-2). Bony prominences such as the ASIS and greater trochanter are

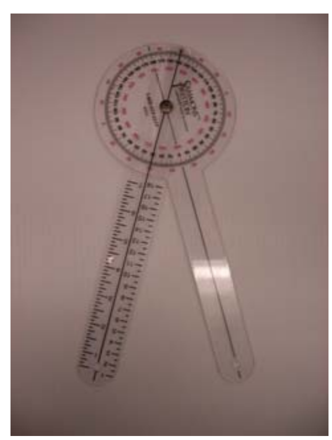


Figure 2-2: A simple 360° goniometer.

used as participant-specific landmarks for either the centre or lever arm of the goniometer. Inter-tester reliability of the MTT has been documented as having an inter-class correlation (ICC) of 0.92 (Gabbe, Bennell, Wajswelner, & Finch, 2004). However, the ICC for test-retest reliability (taken 1 week apart) ranged between 0.63 and 0.75, suggesting hip joint measurements may vary slightly over time.. Fixating a spirit level to one of the arms of the goniometer will enhance the accuracy, when measuring in reference to the vertical (Gabbe et al., 2004).

The following hip measurements are typically done in the supine position:

- Flexion: the centre of the goniometer is placed over the greater trochanter, with the lever arms following the line of the spine, and the femur. Other methods involve using the vertical or horizontal as a reference to the line of the femur.
- Abduction/adduction: centre of goniometer placed over the hip joint (slightly medial to the ASIS) of the leg being measured, with the lever arms in line with the opposite ASIS and the line of the femur.

Hip extension is usually measured by the modified Thomas test (MTT). This procedure, also done in supine, requires the participant to be positioned on the plinth, or table, with their buttocks at the very end of the support surface. The opposite hip is flexed to a degree where the lumbar spine is in a neutral position. This position is maintained either with the subject’s arms, or by the examiner applying pressure to the flexed leg, while the leg of interest is allowed to hang free into extension, with ab/adduction being controlled by the investigator. In this position, the centre of the goniometer is placed atop the greater trochanter on the lateral thigh, with the lever arms following the lateral line of the femur and the vertical. With the knee relaxed, this measurement represents flexibility of iliacus and psoas: the major hip flexors that cross the hip joint but not the knee. Flexibility of rectus femoris, which crosses both the knee and the hip joints, can be further assessed in this position by measuring the knee flexion angle, or observing the effect that further knee flexion has on the angle of hip joint flexion (Gabbe et al., 2004; D. Harvey, 1998). Similarly, abduction of the leg must be controlled to understand the effect that tensor fascia latae (TFL) has on the hanging limb: tightness in this structure will tend to abduct the hip, especially so in slight knee flexion. But the investigator is cautioned to not make assumptions with regard to passive measurements correlating with dynamic joint movement: the modified Thomas test has been shown to correlate poorly with the amount of hip extension demonstrated in active gait (R. Y. W. Lee & Wong, 2002; Schache et al., 2000).

Hip rotation is measured in various positions in the literature: supine with hip and knee flexion, sitting with knees at the edge of a chair/plinth, supine with knees flexed over the end of plinth (but hips in a neutral/extended position), or prone with knee flexion. The latter method appears to be the most appropriate for a study involving gait-type mechanics, as it allows the hip to be in extension, thus not putting increased tension on the posterior hip capsule/soft tissues as would happen in a position of 90° flexion. There often exists significant variability when comparing prone and supine hip rotation measurements (Simoneau et al., 1998), which is likely explained by a change in tension in the anterior and posterior hip structures. Consequently, measurement position must be consistent and documented.

2.11. Elliptical trainer

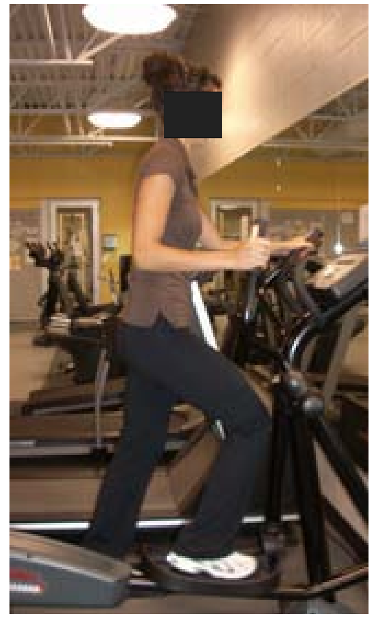


Figure 2-3: An example of a person using the elliptical trainer: both hands using the handles in this case.

The elliptical trainer has gained popularity in recent years due to its relatively low impact requirements, with a metabolic cost comparable to treadmill running (Mier et al., 2006). The elliptical trainer is used in a standing position, with the feet atop footpads that move in an elliptical sagittal path (Figure 2-3). At the same time, the arms move in a contralateral fashion to the legs, pulling two handgrips. Arm involvement is optional: users may choose to use the handgrips, hold onto a stationary bar at the front of the apparatus, or allow their arms to swing freely. Using an elliptical trainer fitted with a 6 component force transducer in one of the pedals, Lu et al (2007) describe significantly greater peak hip, knee and ankle flexion angles on the elliptical when compared to walking, as well as larger peak hip flexor moments. Similarly, an increase in trunk, hip and knee flexion angle and anterior pelvic tilt were described by Burnfield et al (2010) when comparing 4 varieties of elliptical trainers with the kinematics involved during walking. Both of these studies analyzed participants who were holding on to the oscillating handles. Anecdotally, many people also choose to either hold onto a central stationary bar, or not hold on at all when using the elliptical. To the best of my knowledge, the effect of different hand positions on the kinematics has not been researched, nor has lumbar motion in any axis other than the sagittal plane.

Clinically, there are many anecdotes of people commenting that they experience back pain during or after using this piece of exercise equipment. Although the sagittal hip angles appear to be less than those measured during sprinting, the addition of resisted arm activity may increase lumbar compression and shear due to the added amounts of trunk muscle activation required for the arm activity. In addition, the dimensions of the elliptical trainer may not be ideal for all users; people with smaller stature (shorter arms and/or legs) may experience larger hip angles and require greater amounts of axial rotation compared to their taller colleagues. On the elliptical trainer, the position of the feet and arms is dictated by the machine: instead of the biomechanics of the activity being modified to fit the athlete, the athlete has to adapt to the dimensions of the equipment. As previously mentioned, women already tend to run with more hip flexion and adduction, as well as increased axial rotation (Hurd et al., 2004; Pollard et al., 2007; Schache et al., 2003). If a female also happens to be short of stature, the amount of rotation necessary for elliptical trainer use may surpass the safety margin for the lumbar spine, likely the most mobile and vulnerable segment of the kinetic chain, since the pelvis and thorax are rotating in opposite directions (Bruijn et al., 2006). Similarly, leg length correlates negatively with hip extension angle in treadmill running (Schache et al., 2000), and a similar effect may well occur with the elliptical trainer: resulting in hip extension requirements nearing the outer safety limits for a shorter person. In turn, this may tend to increase anterior pelvic tilt and lumbar lordosis. This combination of extreme position and increased muscle activation patterns from handgrip use could result in abnormally high stresses to the lumbar spine.

The other question arising with regards to the elliptical trainer, is what happens if the hip joint is lacking the necessary extension and/or rotation? Since the amount of motion is pre-determined by the equipment and individual anthropometrics, an inability to adequately extend or rotate the hip will no doubt defer the movements to other joints: possibly the knee or the lumbar spine.

2.12. Stretching

Soft tissue joint restrictions have traditionally been associated with an increased incidence of injury (Verrall G.M. et al., 2007), thus stretching is routinely used in rehabilitation settings to increase the available range of motion. The main types of stretching fall into three categories: ballistic, passive, and contract/relax. Ballistic stretching, comprised of quick bouncing movements, has been shown to be the least effective of the three, resulting in a significantly less increase in flexibility than the passive stretch (Wallin, Ekblom, Grahn, & Nordenborg, 1985). Although ballistic exercises may be useful to improve reflex activities and power, the sudden eccentric contractions also put a large tensile stress on the musculo-tendinous (MT) unit, subjecting it to greater risk of damage (Bandy & Irion, 1994; D. C. Taylor, Dalton, Seaber, & Garrett, 1990).

Contract/relax (commonly referred to as proprioceptive neuromuscular facilitation, or PNF) is composed of 3 stages: the tissue to be stretched is taken to near the end of its available range of motion. A maximum isometric contraction of the targeted muscle is then carried out for 7 – 8 seconds, followed by 2 – 3 seconds of rest. Following this, the muscle is once again passively stretched to the limit of its physiological range. Although the literature supports the effectiveness and even superiority of this type of stretching (Bandy & Irion, 1994; Möller, Ekstrand, Öberg, & Gillquist, 1985; Prentice, 1983; Sady, Wortmann, & Blanke, 1982), it generally requires an experienced therapist/partner to administer the passive stretch without overstretching (Weldon & Hill, 2003), thus hindering the ability of participants to be independent with a home stretching program.

Consequently, most stretching programs that encourage participant independence resort to using passive stretch. Many studies have examined the optimum duration that a stretch should be held. It appears that stretching longer than 30 seconds is not necessary; that maximum length change occurs by 30 seconds, with no significant additional increase in flexibility by 60 seconds (Bandy & Irion, 1994; Bandy, Irion, & Baltz, 1994; Bandy, Irion, & Briggler, 1997; Madding, Wong, Hallum, & Medeiros, 1987). This is in keeping with in-vitro studies which show that the maximum creep of a MT unit occurred during the initial 12 – 18 seconds of a slow, static stretch (D. C. Taylor et al., 1990). However, a 60 second stretch has been shown to be more beneficial in people over the age of 65, resulting in a greater total gain in range of motion in less time (Feland, Myrer, Schulthies, Fellingham, & Measom, 2001). This may be due to changes in the muscle stiffness and collagen composition associated with aging. In terms of frequency, once a day over a 30 day period has been shown to be as effective as 3 times per day (stretching 3 - 5 day per week) (Bandy et al., 1997; Wallin et al., 1985). Following a passive stretch routine, measured active range of motion increases more than passive range (Roberts & Wilson, 1993) possibly due to the reciprocal inhibition effect of a contracting muscle inhibiting its antagonist (Prentice, 1983). There is also some suggestion in the literature that stretching applied to muscles with limited extensibility may be more effective than that applied to those with a “normal” length (L. Harvey, Herbert, & Crosbie, 2002; McCue, 1953).

How, exactly, does a muscle/tendon unit change length? Previously, it had been thought to be due to a lessening in the muscle activation of the stretched muscle. However, numerous studies have de-bunked this theory, finding that no significant changes in the amplitude or frequency domain in the low level of EMG activity occur during or after a static stretch (Magnusson, Simonsen, Aagaard, Gleim, & Kjaer, 1995; Magnusson et al., 1996; Magnusson, Simonsen, Aagaard, Sorensen, & Kjaer, 1996; Magnusson, 1998). Short-term changes are mainly a function of the visco-elastic properties of the tissues, thus sensitive to rate and magnitude of the stress applied, as well as the number of repetitions. In-vitro work indicates that significant decreases in peak tension occurs only in the first 4 repetitions, in a series of 10 sequential stretches to the same pre-determined length (Garrett, 1990; D. C. Taylor et al., 1990). When repetitively taken to the same tension, 80% of the length increase occurred during the first 4 stretches. Also demonstrated by these authors was the dependency of peak tensile force and absorbed energy on the stretch rate: a slower rate allows for a greater amount of stress relaxation to occur, thus lower tensile forces (Taylor et al., 1990). Again, this reaffirms the belief that most of the early changes are due to the visco-elastic properties of the MT unit (Magnusson, Aagaard, Simonsen, & Bojsen-Moller, 2000) and adds to the rationale of why slow static stretching is preferable to a ballistic style. One hour after stretching, however, most of the visco-elastic changes (muscle stiffness and stress relaxation) have reversed, and the tissue properties return to a baseline level (Magnusson, Simonsen, Aagaard, & Kjaer, 1996; Magnusson, Simonsen, Aagaard, Sorensen et al., 1996).

As to the different sections of the MT unit: under tension, the tendon has been shown to exhibit 2% strain compared to 8% at the MT junction, indicating the different material properties of these sections (Lieber, Leonard, Brown, & Trestik, 1991). Thus, in a compliant system, some of the tensile force will be attenuated at the MT junction. A stiffer segment, though, will transfer more of the force to the contractile tissue, putting it at greater risk of injury (Weldon & Hill, 2003). In animal models, the number of serial sarcomeres begins to change within 24 hours in an immobilized muscle, so as to optimize the available force, based on the length of the MT unit. That is, when stretched, the number of sarcomeres increases and the length of each one decreases, so as to maintain the optimum length of each sarcomere with regards to force production. Conversely, immobilization in a shortened position will result in a decrease in the number (and increase in length) of serial sarcomeres (Gossman, Sahrmann, & Rose, 1982; Williams & Goldspink, 1973; Williams, 1990). A muscle which has adapted to a shorter resting length is less able to tolerate active and passive tension than one of regular length, possibly due to a relative increase in the amount of connective tissue, which changes properties at a slower rate, resulting in less extensibility of the MT unit (Gossman et al., 1982).

There is a poverty of information in the literature explaining the effects of a long term stretching program in humans. It appears obvious by observing gymnasts, dancers, and unilaterally dominant sports such as baseball and tennis, that a stretching routine over many months and years can alter flexibility (Ellenbecker et al., 2007). However, research suggests that, especially in the short term, much of the observed change may be due more so to an increase in stretch tolerance than an actual lengthening of the structure. Changes in force measured at the same angle of knee extension when stretching the hamstrings, or an increase in length with the same applied force, would be expected with a true increase in soft tissue length. However, after a 3 week hamstring stretching program, neither of these was demonstrated, despite a significant increase in knee angle, indicating personal tolerance to stretch may be responsible for the change in total range of motion rather than tissue properties (Björklund, Hamberg, & Crenshaw, 2001; Folpp, Deall, Harvey, & Gwinn, 2006; Magnusson, Simonsen, Aagaard, Sorensen et al., 1996; Magnusson, 1998). Similarly, participants who demonstrated decreased flexibility on a toe-touch test were found to have not only stiffer hamstring muscles, but also a lower tolerance to passive stretch (Magnusson, 1998).

However, the bottom line is that, whether it is due to personal stretch tolerance or a true increased length in the restricting soft tissues, many studies have found a significant increase in measurable range of motion after 4-6 weeks of a regular stretching routine (Bandy & Irion, 1994; Bandy et al., 1997; Feland et al., 2001; Folpp et al., 2006; Gajdosik, 1991; Kuukkanen & Mälkiä, 2000; Martin et al., 1997; Roberts & Wilson, 1993; Thacker, Gilchrist, Stroup, & Kimsey, 2004; Wallin et al., 1985). But it must be remembered that these numbers refer to the degree of flexibility the participant is willing to tolerate, and may or may not be due to a true change in MT unit length. In addition, the amount of flexibility gained during training is easily lost over the next months if the exercise routine is not maintained (Kuukkanen & Mälkiä, 2000).

3. Methodological Considerations

3.1. Model Validation

Previous data collections in this laboratory have used an electromagnetic tracking device, the 3-Space ISOTRAK (Polhemus Inc., Colchester, USA), for collecting and calculating joint angles (Brown & McGill, 2008; Callaghan & McGill, 1995; Kavcic, Grenier, & McGill, 2004a; Moreside, Vera-Garcia, & McGill, 2006). However, two of the data collections in this thesis were utilizing the Vicon MX Motion System (Oxford, UK) to collect marker coordinate data from a series of reflective markers attached to body segments via rigid plates. These data were then processed with software from Visual 3D (C-Motion, Kingston, Canada) to calculate joint angles. Thus, it was necessary to validate the 3-dimensional hip and spine angles calculated using the Vicon/V3D system by comparing them to those of the 3-Space.

3.1.1. Methods

A data collection took place in which a person was instrumented with the two different systems at the same time. The electromagnetic source for the 3-Space was attached to the posterior pelvis, with sensors attached to the distal lateral thigh and over the T12 spinous process. At the same time, reflective markers were applied in the same configuration as that being used for data collection in chapters 5, 6 and 9. Marker coordinate data was collected using the Vicon MX Motion System and processed in Visual 3D (C-Motion). Thus, both systems were concurrently recording the same motions. The participant was asked to flex, bend, and twist the lumbar spine sequentially in one trial (encouraging off-axis movements), allowing the associated hip motion that would naturally occur with such movements. Pearson correlations were conducted to investigate the linear dependence of the two outcomes.

3.1.2. Results

The subsequent spine and hip angles calculated in both systems appear in Figure 3-1.

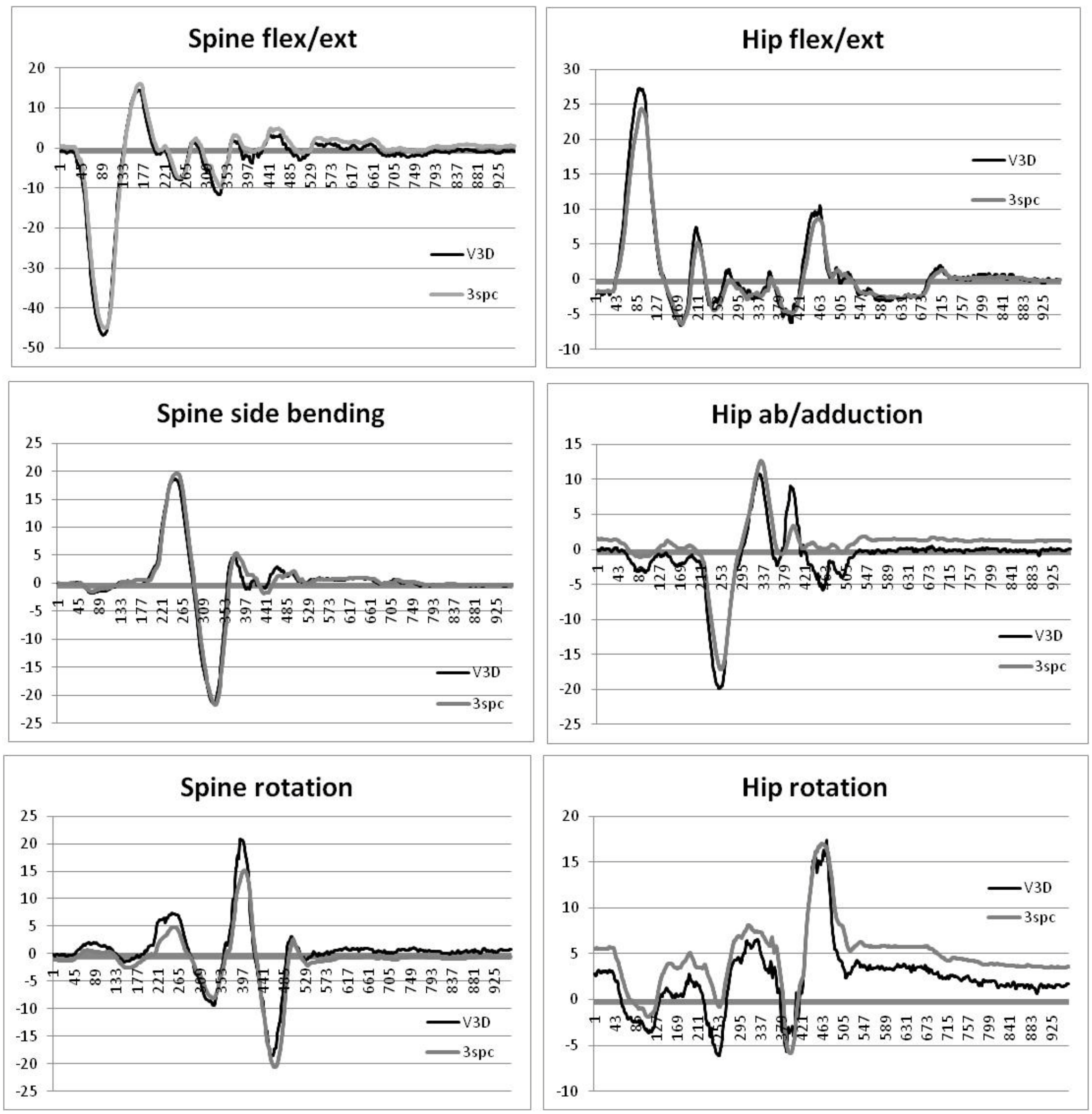


Figure 3-1: Overlay of data collected simultaneously using two methods: the 3-space Isotrak electromagnetic system, and the Vicon MX motion capture system, with processing in Visual 3D

As can be seen in Figure 3-1 and Table 3-1, the two systems react in a nearly identical manner for tracking movement in the spine and for hip flexion. Although still strong, the correlations are slightly less so for hip abduction and rotation, possibly due to the increased soft tissue artifact that is associated with thigh motion and/or the issue of rotation being the third of the angles to be calculated in a Cardan sequence, thus accumulating the errors from the previous calculations (Benoit et al., 2006; Cappozzo, Catani, Leardini, Benedetti, & Della Croce, 1996; Davis, Ounpuu, Tyburski, & Gage, 1991). Both systems were processed using a decomposition order of flexion/extension, ab/adduction (usually referred to as side bending in the case of the spine), then axial rotation.

Table 3-1: Pearson correlations comparing angles calculated using 3 space Isotrak, and those collected and processed using Vicon motion capture and Visual 3D.

	Flexion/extension	Side bending	Twist
spine	0.984	0.968	0.930
hip	0.976	0.886	0.905

3.1.3. Summary

In that the main focus of these studies was spine motion and hip flexion/extension, collecting and processing data using the Vicon and Visual 3D was deemed valid, when compared to the alternate 3 Space system.

4. Quantifying normal hip ROM in healthy young adult males

4.1. Introduction

To understand the effect that hip mobility has on the lumbar spine, one must first understand what constitutes normal range of motion (ROM). The literature quantifying hip ROM is mixed. Measurement outcomes will be affected by the position of the participant (Moreside & McGill, submitted; Simoneau et al., 1998) and whether the end ROM is obtained actively or passively (James & Parker, 1989). Hip extension has been shown to significantly decrease with age (Kerrigan et al., 2001; Roach & Miles, 1991), both in passive measurements and during gait, but no significant changes have been demonstrated in the other directions (flexion, ab/adduction or rotation). Confusion exists as to discrepancies between the sexes. While some authors describe increased external rotation (ER) in females (Roach & Miles, 1991; Simoneau et al., 1998), others have found the opposite: more in the males (Gombatto et al., 2006; Mellin, 1990). Increased extension, abduction and internal rotation in females has also been described by Bach et al (1985) and Mellin (1990), with the latter using a low back pain population.

To narrow down these variabilities, it was decided to constrain the studies to a young, adult male population (age 18 – 35). A review of the literature regarding normal ROM in this demographic demonstrates the variability that exists, depending on whether the measurements were obtained actively or passively, as well as the position of each participant (Table 4-1).

Table 4-1: Average(SD) range of hip joint rotation and extension for a young adult male population, as published in the literature. * no total rotation numbers were published: this is simply an addition of IR and ER, thus no SD is available. ¥ supine measurement was obtained with the hip joint in neutral flexion/extension, and the knee flexed over the end of the plinthe.

Authors	**Active/ passive**	**Age (years)**	**IR**	**ER**	**TotR**	**Ext**
Simoneau et al (1998)	A	18 - 26	32(9)° prone 30(7) ° seated	44(7) ° Prone 35(8) ° seated	76(10) ° Prone 65(8) ° seated	______ ______
Roach and Miles (1991)	A	25 - 39	33(7) ° seated	34(8) ° seated	77(*) seated	22(8) ° prone
Manning and Hudson (2009)	P	26(3.5)	25(6) ° supine	44(3) ° supine	69(*) supine	17.5(2) ° prone
Malliaris et al (2009)	P	15 - 21	34(11) ° prone	48(10) ° supine¥	82(*)° mixed	______

While the reported literature often refer to the seated and prone positions as being those most commonly used in a clinic, this investigator's experience is that the preferred positions for measuring passive hip rotation are supine, with the knee and hip both flexed to 90° (0° ab/adduction) or prone with the hip in neutral and knee flexed to 90°.

Passive hip extension is usually measured using the Modified Thomas Test (MTT), with the participants in a supine position with the leg of interest over the end of the plinth (Kendall & McCreary, 1983). Inter-tester reliability of the MTT has been documented as having an inter-class correlation (ICC) of 0.92 (Gabbe et al., 2004). However, the ICC for test-retest reliability (taken 1 week apart) ranged between 0.63 and 0.75, suggesting hip joint measurements may vary slightly over time.

Furthermore, previous data collections (Table 4-1) have used either a goniometer or inclinometer to measure hip rotation and extension. The reliability of goniometric measurements has been documented: ICC's of 0.94, 0.88 and 0.92 were reported for hip IR, ER and extension respectively (Gabbe et al., 2004). Inter-tester variability has been shown to be higher than intra-tester, thus limiting measuring to a single investigator is recommended (Boone et al., 1978; Bovens, van Baak, Vrencken, Wijnen, & Verstappen F.T.J., 1990). Much of the error associated with using a goniometer has been attributed to faulty application (Ekstrand, Wiktorsson, Öberg, & Gillquist, 1982). It was therefore suggested that measuring body segment positions with a motion capture system, such as the Vicon MX Motion System , would result in more objective outcome measurements, as application error would be minimized.

Of the above listed articles, only Roach & Miles (1991) published percentile data (25th and 75th) along with the averages. In that the goal of this thesis was to compare males with limited hip mobility to those with normal or excessive hip mobility, it was important that normative data was collected to define how much motion would be representative of these two groups.

The purpose of this study was to collect normative data that would represent available passive hip IR, ER and extension in a young (18 – 35) male population. The intention was to both unify the previous literature and to adapt methods to address some of the limitations. Thus, the measurements in this study are to be obtained using a Vicon MX motion capture system, and with the same investigator positioning every participant, to minimize errors due to intra-tester variability. Furthermore, to enhance the application of the data of this study to track hip motion variability, percentile scores were calculated. A sub-study was also conducted over 22 participants to compare measurements obtained with the Vicon MX system to those measured with a goniometer.

4.2. Methods

Young men were recruited for this study. Their hip internal rotation (IR), ER and extension passive ROM was measured using the Vicon MX system for motion capture, and subsequent angles calculated mathematically. Twenty-two of the participants also had ROM measured concurrently with a goniometer.

4.2.1. Participants

In total, 77 participants were recruited from the university setting and the local community, via posters and word of mouth (mean age = 22.8 (3.2) years; mean height = 179.7(6.6) cm; mean mass = 78.9(12.0) kg). Sixty-eight of the seventy-seven claimed that their right leg was dominant, as was decided by which was the preferred leg for kicking a ball. All subjects were healthy without current hip or back pain or past pathology in these regions. Participants completed a written informed consent document approved by the University of Waterloo Office for Research Ethics.

4.2.2. Hip Range of Motion Measurements

To facilitate quantitative measures of motion using an infrared based 3D motion capture system (Vicon MX Motion System), three non-colinear reflective markers were placed bilaterally on the thigh and shin, using the following landmarks: tibial tubercle, anterior mid-shin at the level of the ankle, lateral mid-shin approximately 1/3 of the way down the shin, greater trochanter of the thigh, lateral femoral condyle, and anterior-lateral thigh, also about 1/3 of the distance down the thigh, distal to the greater trochanter. The pelvis markers were attached lateral to the ASIS's (to allow prone postures without dislodging them, and improved visibility when the hip was flexed), as well as two additional markers superior to the right pelvis marker, one approximately 4 cm in the anterior orientation, one the same distance posterior. The investigator was responsible for placement of all reflective markers, as well as passive positioning of each participant, and has over 30 years experience as a clinical physiotherapist. Data were collected in static positions and lasting approximately 3 seconds. Order of collection trials in each position was randomized, as was the variable of supine or prone, and right or left hip. Each measurement was repeated twice. In that the measurements were calculated mathematically after the trials, there was no immediate feedback to indicate position at the time of capture.

Initial supine quiet lying trials were captured with the participant in relaxed supine lying, and feet passively oriented in a vertical direction by the investigator, to standardize for neutral thigh/leg position. Supine hip extension was measured using the Modified Thomas Test, with the investigator controlling for ab/adduction and rotation (Figure 4-1). However, it was decided after the first 11 participants that the simple MTT did not allow for enough objective control of the pelvis and lumbar spine. Subsequently, a blood pressure (BP) cuff was placed under the lumbar spine: the investigator would flex both hips/knees while maintaining her hand under the lumbar lordosis, until she and the participant both agreed that the lordosis had reduced to a more neutral position, indicating posterior rotation of the pelvis in the sagittal plane. The hand was then removed from the low back,

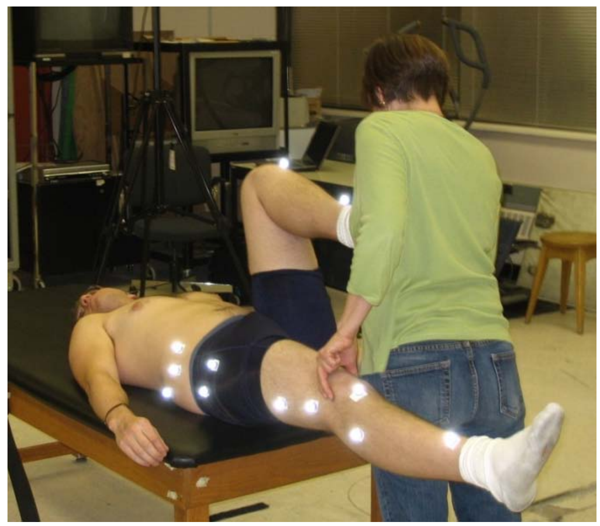


Figure 4-1: Position adopted during the Modified Thomas Test, to measure hip extension. Here, the right thigh is being slowly lowered to full extension, while controlling for ab/adduction. The investigator maintains the right hip in passive flexion to minimize sagittal rotation of the pelvis.

and a blood pressure cuff replaced it, with the cuff then being inflated to 60 mmHg. This pressure was monitored as one of the participant's legs was lowered passively to a position of maximum hip extension without associated changes in pelvic position/pressure in the BP cuff. The opposite leg was held passively in a position of hip/knee flexion by the investigator, which maintained the BP cuff at approximately 60 mm Hg. Participants were encouraged to give feedback as to their perception of pelvis position, in an attempt to further minimize pelvic rotation during hip extended.

Figure 4-2: The three positions used for measuring prone hip rotations: neutral quiet lying (a), bilateral internal hip rotation (b) and internal rotation, here shown for the left hip (c).

Hip rotation measurements were captured with participants in a prone posture. An initial quiet lying posture required both knees to be bent approximately 90°, with the shanks passively oriented vertically towards the ceiling by the investigator (Figure 4-2a). These quiet lying trials became the data from which a "bias", was later subtracted from the rotation trials, to remove error induced by marker placement. Bilateral internal rotation measurements were done simultaneously, as participants were asked to let both lower legs fall out to the side, while maintaining the knees at 90° of bend (Figure 4-2b). External rotation required the leg of interest to passively rotate across the midline (Figure 4-2c). Pressure was applied on the ipsilateral pelvis by the investigator to ensure pelvis rotation did not occur. In those cases where large amounts of hip ER was present, the non-tested leg was abducted approximately 10° to allow free motion of the tested leg (Barbee Ellison et al., 1990).

Twenty-two of the participants also had hip extension and prone rotation measured manually with a goniometer at the same time as the Vicon captures, by a visiting physiotherapist. In that he was only able to take part in the study for 2 weeks, the number of participants measured was limited to 22. The goniometer was modified with the addition of two spirit levels: one on each of the arms, to improve accuracy of determining horizontal and vertical positioning (Figure 4-3) (Gabbe et al., 2004). All measurements were done by this one therapist, who subsequently passed the goniometer to another person to read and record the angle. Thus, the principle investigator and visiting physiotherapist were blinded to the measurement.

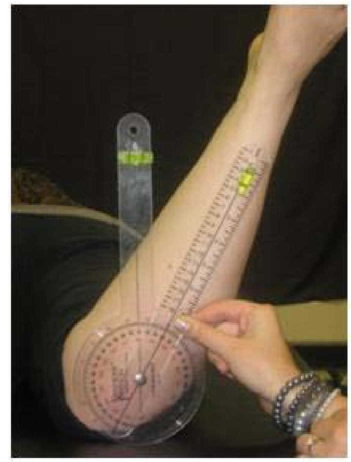


Figure 4-3: Measurement of right hip internal rotation with a goniometer affixed with two spirit levels, to improve accuracy of aligning with the vertical or horizontal.

4.2.3. <u>Calculation of angles</u>

Since the collected Vicon data was of a 3 second static trial, an average was taken of the marker placement over each capture. This average was used to represent marker position for ongoing calculations.

Supine hip extension angles were originally calculated using Euler angles, with the angle of the thigh being calculated relative to the pelvis. However, it soon became apparent that the error associated with the pelvis markers created a large signal to noise ratio, most likely due to skin movement artifact. For example, when the right knee was flexed near to or past 90°, the three non-colinear markers on the right pelvis moved relative to the underlying bony pelvis, as the skin in that region buckled. In that the participants were lying supine or prone for all captures, standard pelvis markers configurations, such as the Helen Hayes or -Coda could not be used, as all require posterior sacral markers.

Thus, the thigh angle for extension was calculated relative to the laboratory coordinate system, knowing that the pelvis position was being monitored with the blood pressure cuff. The laboratory system was set up such that the z axis was vertical, the y axis was oriented across the plinthe (treatment table), and the x axis was in line with the length of the plinth. Angle calculations were done twice: the first using simple 2d angles:

$$Angle = \arctan [(DTz - PTz)/(DTx - PTx)]$$

(DT = distal thigh, PT = proximal thigh, x and y indicate the axis in a 3d coordinate system). 3d angles were also calculated using an Euler angle method utilizing Mathcad software (PTC, Needham, USA).

Prone rotation angles were calculated relative to the vertical z-axis, using the following equation:

$$Angle = \arctan[(DSy-PSy)/(DSz-PSz)]$$

(DS = distal shin, PT = proximal shin, y and z indicate the axis in a 3d coordinate system).

For each participant, the angles calculated in the quiet lying trials were then subtracted from each subsequent trial as a bias angle, to accommodate for marker positioning differences, tibial bowing, etc. For each leg, angles of IR and ER were added together, resulting in a total hip rotation measurement (TRot), which was subsequently used in the normative and percentile data calculations.

4.2.4. Statistical Analysis

All analyses utilized the SPSS (version 17) package with a significance level chosen at $p < 0.05$. Pearson correlations were performed to compare hip extension measurements calculated in 2d and 3d. For each hip angle measurement type (Ext, IR, ER, TRot), paired t-tests with Bonferroni adjustments were calculated to compare right and left sides, resulting in a significance level of 0.0125 (0.05/4). If no significant differences were found, these right and left data were collapsed. Q-Q plots and Shapiro-Wilk tests were then used to test normal distribution. From these, angles were determined that represent hip extension and rotation measurements at relevant percentiles. In a separate analysis, angles obtained from the goniometric measurements of 22 participants were compared to those calculated using the Vicon data. Again, paired t-tests were used to compare right and left sides, which were collapsed if not different. T-tests with Bonferroni adjustments were then used to compare goniometer to Vicon (Ext, IR, ER, and TRot). Pearson correlations were conducted comparing the ROM of total hip rotation with hip extension (Rt and Lt sides collapsed).

4.3. Results

4.3.1. Hip extension

Pearson correlations between 2d and 3d hip extension angles for both the right and left sides were 0.99 with r^2 values of 0.975 and 0.98, respectively (Figure 4-5). This high association indicates minimal difference between the two computational methods, and 2d was subsequently chosen for ongoing analyses.

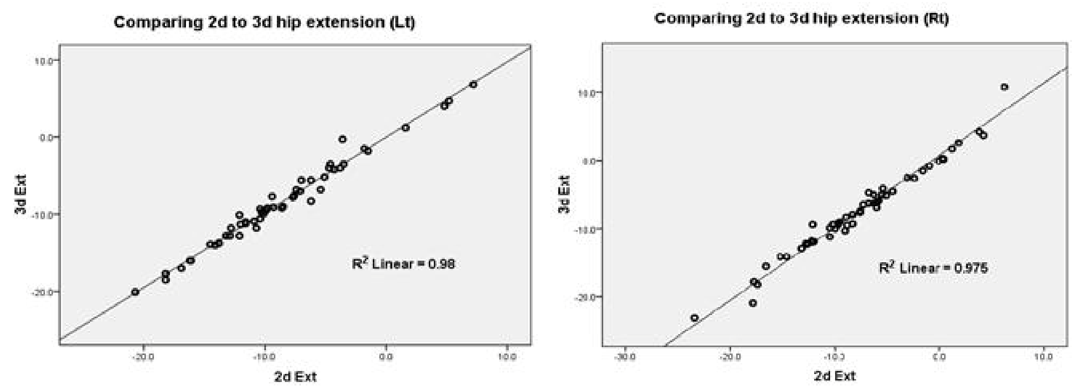


Figure 4-5: Correlations of hip extension measurements calculated in 2d and 3d, for the left and right sides. (n = 53 and 51 for Rt and Lt, respectively)

Paired t-tests indicated no significant difference between right and left hip extension (p = 0.213, n = 61 per group) (Table 4-4). Right and left sides were thus collapsed for normative data analyses. The Q-Q plot in Figure 4-4 plots hip extension data against hypothetical data that would represent a perfectly normal distribution. As seen in the graph, the data deviates very little from normal, and no significant difference is found using the Shapiro-Wilk test for normality (p = 0.609*)*. Percentiles were then calculated which represent the pertinent hip extension ROM representative of this data set, as shown in Table 4-2. To clarify, the 5^{th} percentile represents the least amount of hip extension, and the 95^{th} the most. A negative number reflects a lack of extension in the MTT, thus the thigh lies above the horizontal. As is standard with the MTT, 10° of hip extension has been added to the measured angles, to account for flexion of the pelvis, and relative hip extension. Despite the fact that

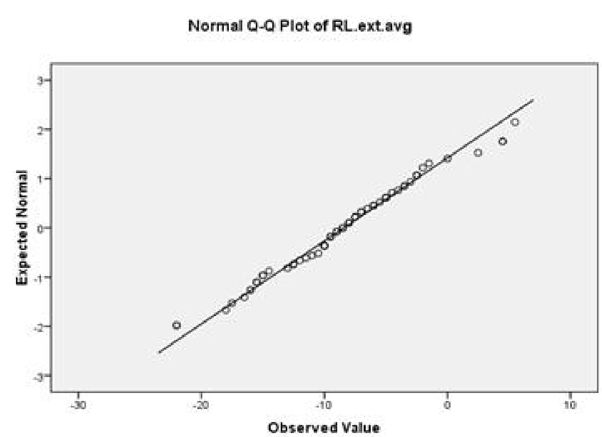


Figure 4-4: Q-Q plot demonstrating the normative distribution of 2d hip extension data (Rt/Lt collapsed). (n = 62)

these numbers may represent the true amount of hip extension relative to the pelvis, they can be confusing when trying to picture the position of the thigh in space. Thus, the bottom row gives the same percentiles in terms of what would be visualized or measured with a goniometer relative to the horizontal (thus, +ve means the thigh is above the horizontal, -ve is below), with no accounting for pelvis flexion.

Table 4-2: Percentile data for supine passive hip extension, where right and left measurements were collapsed. 50th percentile includes standard deviation: mean (SD). Calculations were made in 2d from data collected with the Vicon MX Motion System.

	Percentiles: Right/Left hip extension, collapsed						
Percentile	5th	10th	25th	50th	75th	90th	95th
Modified Thomas Test (MTT)	-18°	-16°	-12°	-8(6)°	-5°	-2°	4°
Relative to horizontal, no correction for pelvis	+8°	+6°	+2°	-2(6)°	-5°	-8°	-14°

4.3.2. <u>Hip rotation</u>

Paired t-tests indicated a statistically significant difference between right and left sides in TRot, with the mean of the left hip being 56.6° compared to 61.4° on the right (mean difference: 2.48, 95% CI: 0.62 – 4.33, df = 66, p = 0.01). Upon closer examination, there were 3 participants who demonstrated a greater than 17° difference between the right and left TRot, which, although interesting and clinically relevant, would be considered highly unusual. It was decided to remove the data from these 3 participants, resulting in a p-value of 0.052, allowing the right/left sides to be collapsed (Table 4-4). The Q-Q plot (Figure 4-6) again shows the

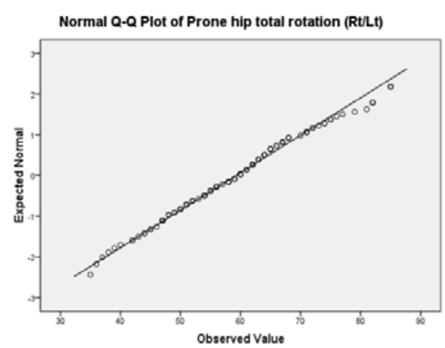


Figure 4-6: Q-Q plot of total hip rotation, demonstrating little difference from a normal distribution. (n = 73)

TRot data closely follows the normal line, and Shapiro-Wilk testing resulted in a significance level of 0.377. The associated percentiles for total hip rotation appear in Table 4-3.

Table 4-3: Percentile data for prone passive hip rotation, where right and left measurements were collapsed. 50th percentile includes standard deviation: mean (SD). Calculations were made in 2d from data collected with the Vicon MX Motion System.

	Percentiles: Hip rotations						
Percentile	5th	10th	25th	50th	75th	90th	95th
Total Rotation	44°	46°	53°	59(11)°	66°	75°	82°
Internal Rotation	12°	15°	20°	26(8)°	31°	37°	42°
External Rotation	19°	23°	28°	34(9)°	40°	46°	50°

Paired t-tests comparing right and left sides for hip IR and ER demonstrated no significant differences between sides (see Table 4-4). Subsequent Shapiro-Wilk tests of the collapsed data showed no significant difference from a normal distribution (p = 0.125, 0.101 for ER and IR, respectively). Percentile data is shown in Table 4-3.

4.3.3. Goniometer

There were no significant differences between the Rt and Lt sides in hip extension or rotation measurements obtained with the goniometer (Table 4-4) thus right and left were collapsed. There was a significant difference, however, between the goniometric measurements of hip extension and those calculated in 2d ($p < 0.001$), with the goniometer measurements being an average of 3.9° less (indicating greater

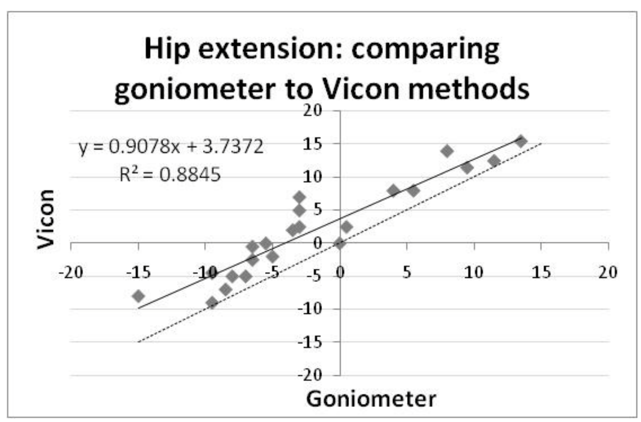


Figure 4-7: Scatterplot of the hip extension measurements obtained with the goniometer compared to the Vicon system. The solid line are the results, the dotted line represents a hypothetical correlation of 1.0, thus demonstrating that the goniometric measurement was consistantly less than the Vicon. (n = 22 per group)

amounts of extension) (95% CI: -5.1- -2.8, df = 21). In spite of this, the Pearson correlation was high (0.940), with an r^2 value of 0.88, indicating high correlation between the two techniques. Figure 4-7 demonstrates the overall tendency of the goniometer to measure less than the passive marker system. Similarly, statistically significant differences exist between the two measurement types for ER (p = 0.046). However, the goniometric measurements were less than one degree different than the Vicon, which would be considered within the margin of error for clinical measurement (mean: -0.92°, 95% CI: -1.8 – -0.16, df = 19). No significant differences were demonstrated between the Vicon and goniometer for IR (mean 0.72°, 95% CI: -0.21 – 1.66, df = 19, p = 0.122).

Table 4-4: Results from paired t-tests comparing Rt vs. Lt of various hip motions and type of measurement.

Comparison	**Mean diff (°)**	**95% CI**	**df**	***p*-value**
Hip ext: 2d	0.82	-0.45 – 2.08	59	0.20
Hip ext: gon	2.1	-0.85 – 5.1	18	0.15
Hip TRot	1.69	-0.015 – 3.39	63	0.05
Hip IR	0.03	-1.53 – 1.59	66	0.97
Hip ER	2.3	0.26 – 4.35	66	0.03
Hip IR gon	0.50	-2.94 – 5.94	19	0.67
Hip ER gon	0.85	-2.94 – 4.69	19	0.64

4.4. Discussion

Accurate normative data is the basis for quality research that analyzes differences between percentile groups, or changes that ensue due to an intervention. The data presented here represent hip extension and prone rotation measurements obtained with the Vicon MX motion capture system, with relevant calculations being done in 2d and 3d. The high correlation between 2d and 3d hip extension measurements allows the researcher/clinician to choose whichever method is more convenient for their specific project. Despite the fact that motion capture systems are likely more accurate, the goniometer remains a viable clinical tool. Thus, knowing that the goniometer correlates highly with the Vicon measurements for hip rotation may encourage data collections in a clinical setting, where motion capture equipment is not readily available. Goniometric measurements of hip extension, however, should be done with the understanding that they tend to overstate extension by an average of 3.9°, when compared to the Vicon system. Researchers should thus look for larger disparities between groups if using the goniometer as their main method of measuring extension.

The main limitation of this study is the method used for positioning the participants to measure hip extension: the modified Thomas test. Over the course of many participants, it became obvious that stabilizing the pelvis to accurately measure hip extension ROM relies heavily on the investigator subjectively determining when the pelvis begins to rotate, and counteract that rotation with increased contra-lateral hip flexion force. Positioning of the BP cuff was helpful to provide objective feedback, but it also required the participant to subjectively indicate if they felt the cuff was midline (as the investigator could not see its specific position). BP cuffs which were laterally displaced would not respond accurately to changes in the lumbar lordosis. However, using motion capture systems to measure the pelvis in supine presents its own problems. Attaching markers to the anterior-lateral pelvis may be acceptable with the hip in limited flexion, but full hip flexion, as was required in this study, resulted in a large amount of skin crimping over the anterior hip/pelvis, which tends to distort marker position (skin mounted or on a fin). Thus, the BP cuff seemed to be the best option, which is easily reproducible in most research or clinical environments. This research is further limited by the specific population that was studied: healthy young males. Future investigations should broaden the participant base to include different age groups and females.

These outcomes are less than those previously described in the literature, for this population base. The amount of end range pressure applied to obtain full passive ROM likely differs between investigators. Simoneau et al (1998) and Roach & Miles (1991) both measured active ROM, with participants being asked to use maximum effort. This would tend to encourage stretching of the soft tissues, possibly resulting in a greater ROM than the technique used in this investigation, where ROM was calculated at the time pelvis motion began, without additional over-pressure. Roach and Miles (1991) chose to position their participants in sitting, which would alter not only joint mechanics, but soft tissue tensions in the surrounding structures. Malliaras et al (2009) examined mobility in a younger population: (15 – 21), which could explain some of the increased range. They used a supine position for hip ER, with the knee at the end of the plinthe: thus the thigh supported, knee flexed to 90°. The opposite leg was also in a position of supported hip/knee flexion, which may have altered the pelvis position, compared to this investigation where the participants were prone. Obviously, different positions result in different outcomes. For example, hip external rotation measured in prone averages 9° less than sitting (Simoneau et al., 1998) and 5° less than supine (Moreside & McGill, submitted). Consistency in positioning and method of measurement is vital.

It became apparent during this study that not all participants fit into the same percentiles for both rotation and extension. For example, one participant demonstrated 75° of hip rotation bilaterally, (90th percentile), yet placed near the 30th percentile for extension, with his thigh being 4° above the horizontal. Similarly, another showed the opposite trend: 46° of hip rotation, but 15° below the horizontal for extension (thus 10th and 95th percentiles, respectively). Although these measurements are indicative of extreme cases, they highlight the variability demonstrated in a group of healthy, pain-free young males. Pearson correlations between the two sets of measurements resulted in a value of 0.331, or an r^2 value of 0.110, indicating a weak correlation between the two.

4.5. Summary

A normative data set has been collected for supine hip extension, using the modified Thomas test, and passive prone lying hip internal and external rotation in a young male population. These positions are commonly used and easily reproduced in clinical and research settings. The goniometer is found to be an acceptable measuring device for these hip motions, when compared to the Vicon MX motion capture system. This link between the laboratory and clinical measurement systems may give individuals confidence to carry out hip research in a much less “high-tech” environment. Percentile data gives insight into what constitutes normal, limited, or excessive hip mobility, facilitating comparisons of these groups in further studies investigating the influence of hip mobility. Unlike the typical capsular pattern of hip restriction generally found in the elderly, limitation of motion in either extension or rotation is not necessarily predictive of a similar restriction in the other direction.

5. The Effect of Hip Mobility on Lumbar Spine Motion and Kinetics

5.1. Introduction

Limitation of hip mobility is known to cause secondary positional changes in the pelvis and lumbar spine (Kerrigan et al., 2003; R. Y. W. Lee & Wong, 2002; Murray, 1967; Perry, 1992; Thurston, 1985). Specifically, lack of hip joint extension during gait may result in an increase in anterior pelvic tilt, as well as increased total sagittal spine motion (Kerrigan et al., 2001; Offierski & Macnab, 1983; Perry, 1992). In turn, lack of hip extension or rotation has been associated with low back pain (Chesworth et al., 1994; Fairbank et al., 1984; Kujala et al., 1992; Kujala et al., 1994; Mellin, 1988; Sjolie, 2004; Van Dillen et al., 2008), as has asymmetry of hip joint mobility (Cibulka et al., 1998). Most of these studies, however, have examined people who demonstrate abnormal hip tightness due to arthritic or neurological dysfunctions, or are presenting with a complaint of low back pain. These groups may also tend to show abnormal movement patterns in the lumbar spine due to the nature of their antilogy; the effects of these disorders are likely not isolated to a single joint. Anecdotally, there exists a group of young adults who demonstrate limited hip mobility of unknown origin, and have no associated pain, arthritis or neurological complications. Studying such a group may give insight into compensatory spine movements that occur when lack of motion is available at the hip joint.

Lumbar angles and forces are known to affect disc degeneration and facet joint compression. Specifically, repetitive flexion and extension have been shown to accelerate disc degeneration (Adams, May, Freeman, Morrison, & Dolan, 2000; Aultman, Scannell, &

McGill, 2005; Callaghan & McGill, 2001; Drake, Aultman, McGill, & Callaghan, 2005; Marshall & McGill, 2010; Tampier, Drake, Callaghan, & McGill, 2007) and increase foramenal pressure (Drake & Callaghan, 2009). The addition of axial torque to these sagittal motions hastens disc degeneration (Drake & Callaghan, 2009; Marshall & McGill, 2010). Consequently, in those cases where hip joint mobility is limited, a compensatory increase in spine motion may also accelerate the onset of low back dysfunction. Subsequent changes in muscle activation patterns, in response to the spine angle adaptations to limited hip mobility, may also affect lumbar compression and shear forces (Cholewicki, McGill, & Norman, 1995; Fenwick, Brown, & McGill, 2009; Granata & Marras, 1993; Marras & Granata, 1997; Marras, Knapik, & Ferguson, 2009; Moreside et al., 2006), also known to affect spine health. Thus, it is important to understand the ramifications of limited hip mobility as it pertains to the lumbar spine.

The purpose of this study was to analyze two groups of young adult males: those with limited hip mobility (LHM), and those with excessive hip mobility (EHM). The intent was to compare the amount of hip and spine motion these groups each used when performing numerous functional activities that required hip joint extension and/or rotation, as well as on the elliptical trainer. Subsequently, the effect of these movement patterns on lumbar forces would be investigated. The hypotheses were:

1. The LHM group will stand with an increased anterior pelvic tilt, and demonstrate less available lumbar extension and rotation in upright standing than the EHM group.
2. The LHM group will tend to extend their lumbar spine more than the EHM group during activities which require active hip extension. Similarly, activities

requiring hip rotation will result in more lumbar spine rotation in the LHM group.

3. Muscle activation patterns will reflect any differences that occur in spine motion between the groups.
4. Lumbar spine loading will differ between the 2 groups in those activities where spine motion differs.

5.2. Methods

Twenty healthy young males participated in this study: 10 in each of the LHM and EHM groups. Participants were asked to demonstrate a variety of functional movements: walking, lunging, twisting, extending their hip, in addition to using the elliptical trainer. During these activities, muscle activity was monitored using surface electromyography (EMG) and motion was captured using the Vicon MX Motion System and Nexus Software (Vicon Motion Systems, Oxford, UK). Hip and lumbar spine angles were calculated using Visual 3D software (C-Motion, Kingston, Canada) as were lumbar moments. EMG, lumbar angles and moments were used to drive a custom spine model, calculating lumbar compressive and shear forces, as well as individual muscle contributions. Comparisons were made between the two groups with regards to joint angles, muscle activation patterns and lumbar forces.

5.2.1. Participant recruitment

Participants were recruited from the university population and surrounding area via posters and word of mouth. All claimed to be healthy without current hip or back pain or past pathology in these regions. Each participant completed a written informed consent document approved by the University of Waterloo Office for Research Ethics.

Previous research in this laboratory analyzed hip extension and prone lying hip rotation in a group of 77 males, between the ages of 19 and 30 to determine normative and percentile data for this age group. Results are shown in Table 5-1.

Table 5-1: Percentile data of hip extension and rotation from a young adult male population. 50th percentile data represents the mean (SD).

	Percentiles: Hip rotation and Extension						
	5th	**10th**	**25th**	**50th**	**75th**	**90th**	**95th**
Total Rotation	44°	46°	53°	59(11)°	66°	75°	82°
Internal Rotation	12°	15°	20°	26(8)°	31°	37°	42°
External Rotation	19°	23°	28°	34(9)°	40°	46°	50°
Hip Extension	+8°	+6°	-2°	-1.5(6)	-5°	-8°	-14°

The criteria for participation in this study was to have hip mobility in both directions (extension and total rotation) that was either above or below the 50th percentile. Of interest, there were many potential participants who demonstrated marked limitation in one direction (ie.extension), but greater than average motion in the other (ie. rotation). This had not been expected, and made it more difficult to find participants who demonstrated marked deviations from the mean in both directions. In total, twenty males between the ages of 19 and 30 participated in the study, 10 in each group. Participants were height matched between the groups, averaging 180.1(7.1) cm and 180.1(7.4) cm in height, with mean mass of 84.6(15.3) kg and 78.0(7.1) for LHM and EHM groups, respectively. Individual anthropometric and range of motion data are shown in Appendix *5-1*. Two in the LHM group claimed that their left leg was dominant, as determined by which they would use to kick a soccer ball; all others were right leg dominant. For the trials involving the elliptical trainer, only 9 participants were in each group, as one had difficulty with coordination, thus the data from him and the height-matched participant in the opposing group were removed.

5.2.2. Hip ROM measurements

Hip extension measurements were collected in supine lying, using the modified Thomas test (MTT) with the investigator controlling for ab/adduction and rotation. The participant lay on his back; his pelvis at the end of the plinthe. With the investigator's hand under the participant's lumbar spine, the investigator would flex the participant's hips/knees until she and the participant both agreed that the lordosis had reduced to a neutral position, indicating posterior rotation of the pelvis in the sagittal plane. The investigator's hand was then removed from the low back, and a blood pressure cuff replaced it, (while returning the spine to the same approximate position) with the cuff then being inflated to 60 mmHg. This pressure was monitored as one of the participant's legs was lowered passively by the investigator to a position of maximum hip extension without associated changes in pelvic position/pressure, as monitored by the BP cuff. The opposite leg was held passively in a position of hip/knee flexion by the investigator, which maintained the BP cuff at approximately 60 mm Hg. Participants were encouraged to give feedback as to their perception of pelvis position, in an attempt to further minimize pelvic rotation during hip extension.

Hip rotation measurements were measured with participants in prone lying. Bilateral internal rotation (IR) measurements were done simultaneously, as participants were asked to let both lower legs fall out to the side, while maintaining the knees at 90° of flexion. External rotation (ER) required the leg of interest to passively rotate across the midline. Pressure was applied on the ipsilateral pelvis by the investigator to ensure pelvis rotation did not occur. In those cases where large amounts of hip ER was present, the non-tested leg was abducted approximately 10° to allow free motion of the tested leg (Barbee Ellison et al., 1990).

Extension and rotation measurements were obtained using a standard goniometer modified with the addition of two spirit levels: one on each of the arms, to improve accuracy of determining horizontal and vertical positioning. Every participant was positioned passively into hip extension and rotation by the principal investigator, who has over 30 years' experience as a clinical physiotherapist. Measurements were then obtained by an assistant.

5.2.3. Electromyography

Surface electromyography signals were collected bilaterally on each subject from the following trunk muscles and locations: rectus abdominis (RA), 3 cm lateral to the umbilicus; external oblique (EO), approximately 15 cm lateral to the umbilicus; internal oblique (IO), halfway between the anterior superior iliac spine of the pelvis and the midline, just superior to the inguinal ligament; latissimus dorsi (LD), lateral to T9 over the muscle belly; erector spinae at T9 and L4 (T9ES and L4ES, respectively), located 5 and 1 cm lateral to each spinous process, Gluteus Maximus (GMax), located over the maximal bulk of the muscle belly, approximately mid-buttock, and Gluteus Medius (GMed), approximately 6 cm caudal to the iliac crest on the posterior-lateral pelvis. Pairs of Ag-AgCl surface electrodes were positioned with an inter-electrode distance of 3 cm. The EMG signals were collected at 2400 Hz, and amplified to produce approximately ± 2.5V. EMG signals were full wave rectified and low pass filtered (low pass Butterworth filter) with a cutoff frequency of 2.5 Hz, and then normalized to maximal voluntary isometric contraction (MVC) amplitudes, using a custom Labview program (National Instruments Corp, Austin, USA). The MVC's were obtained during isometric maximal exertion tasks in the following way: for the abdominal muscles, each subject was in a sit up position and manually restrained by a research assistant, who matched the effort so that very little motion occurred. The subject produced a sequence of maximal isometric efforts in trunk flexion, right lateral bend, left lateral bend, right twist and left twist directions, but again with little motion occurring. For the extensor muscles and GMax, an isometric trunk extension was performed with the torso cantilevered over the end of the test table (Biering-Sorensen position). The MVC for GMed was measured with subjects positioned in side lying; the uppermost leg was abducted and slightly externally rotated, with a research assistant resisted maximal isometric efforts of this position. The LD MVC was performed by resisting shoulder adduction and IR at 90 degrees in the frontal plane, although in many instances the maximal activity occurred in the same MVC as the back extensors, thus was chosen for normalization.

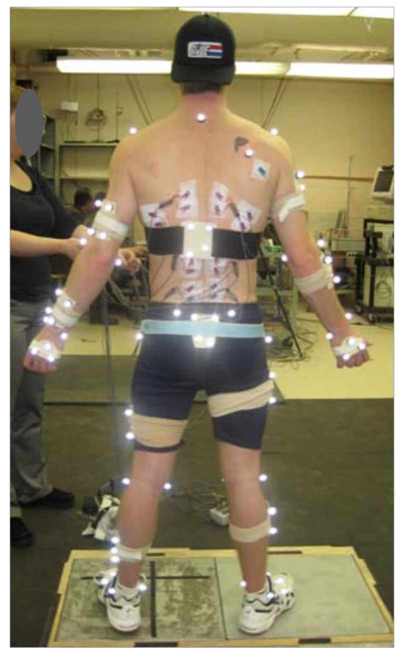


Figure 5-1: Posterior view of participant with reflective markers and EMG electrodes in place.

5.2.4. Motion Capture

Body kinematics were collected during the various functional activities using the Vicon MX Motion System and Nexus software, using eight infra-red cameras collecting at a frequency of 60 Hz. Rigid plates with 4 reflective markers on each were attached via elastic straps to body segments bilaterally as follows: shin, thigh, foot, hand, forearm, upper arm, and overlying the midline on the posterior pelvis, T12 and forehead. In addition, single markers for calibration purposes only were attached over the posterior Rt scapula, C7 spinous process, sternal notch, and bilaterally over the medial and lateral aspects of each ankle, knee, wrist, elbow, ASISs, PSISs, greater trochanters, acromions, and earlobes (Figure 5-1).

5.2.5. Elliptical trainer

An Octane (Octane Fitness, Brooklyn Park, MN USA) elliptical trainer was used for this research, as it featured variable stride lengths and was felt to represent the type of equipment commonly found in a fitness facility. The arms and handles of the elliptical trainer were replaced with ones fabricated from rolled steel, which included triaxial force cubes (AMTI, Watertown, MA, USA) bolted between the upright arms and the handles (Figure 5-2). The forces and moments from these were collected at 2400 Hz, using the Vicon MX software, then further processed using Visual 3D software. Reflective markers were attached to the side of the force cube, as well as the top and bottom of the handle, to track handle and force cube position.

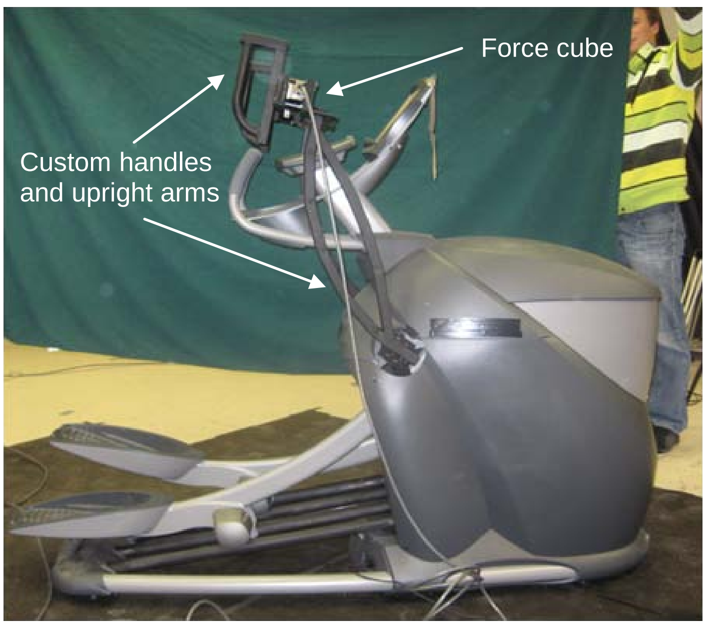


Figure 5-2: Octane elliptical trainer with custom handles and tri-axial force transducer.

5.2.6. Collection procedure

After being initially screened for hip mobility appropriate for the study, participants were scheduled to return for their initial intake session. Upon their return, anthropometric measurements were taken (height, mass, leg, shank and arm lengths), and hip ROM measurements were repeated. Participants were encouraged to practice using the elliptical trainer at this point in time, to ensure comfort and coordination with its use. Surface EMG electrodes were then attached and relevant MVCs collected. Reflective infra-red markers were attached to the body as described above, following which a calibration pose was collected with the motion capture system. Calibration markers were then removed.

Motion capture began with the participants being asked to walk at a self-selected comfortable pace along the length of the laboratory. A force plate was not used, as pilot testing had indicated that constraining the foot position to a specific target resulted in abnormal walking patterns (stride length and cadence changes). Resulting walking speeds ranged from 41 – 60 cycles per minute (cpm), with the mean being 51.1(4) cpm. Stride length ranged from 56 – 90 cm, with a mean of 73.7(8) cm (22 – 36 inches, mean 29(3) inches, for comparison with elliptical specifications). Active lumbar ROM was also collected for flexion, extension and rotation. For the "hip extension" trial, participants were asked to actively extend their hip to their perceived maximum range, while in an upright standing posture. They were given minimal guidance as to how to perform the action, other than to attempt to keep their upper body erect (i.e. avoid leaning the trunk forwards). Hip extension was performed twice on each leg. Next, they were asked to complete a forward lunge: from their standing position, the floor was marked at a distance 1.5X their shin length. They were asked to step forward with one foot, until their toe reached the floor marking, and lunge down into forward hip flexion (the backward hip being extended) as low as was comfortable, while keeping their upper body erect. Again, this was repeated twice on each leg. The last functional trial was a "twist and reach" activity: two poles were set up aside the participants (in the frontal plane). The distance between the poles was 110% their body height, with the participant standing in the middle, feet shoulder width apart. Small knobs on the poles were secured at approximately the height of the person's waist. They were asked to reach around and touch the knob on the right pole with their left hand, and the left pole with the right, while not moving their feet. All trials were repeated twice.

Participants were subsequently asked to use the elliptical trainer. They chose a self-selected speed, with the instructions being, "choose a speed you would feel comfortable using for 30 minutes if exercising in the gym", resulting in velocities between 40 and 70 cycles per minute, with a mean speed of 53(7). During the collection, speed was varied between this self-selected one, and a velocity that was 30% faster (hereafter referred to as "normal" and "fast"). Stride length was varied between 2 positions: 18 inch stride and 26 inch (46 cm and 66 cm, respectively) stride, those being the maximum and minimum stride length available for this specific brand of elliptical trainer. Hand position varied between 3 options (Figure 5-3):

1. Holding onto the handles ("handles")
2. Holding onto the central support bar ("bar")
3. Not holding on to anything ("freehand")

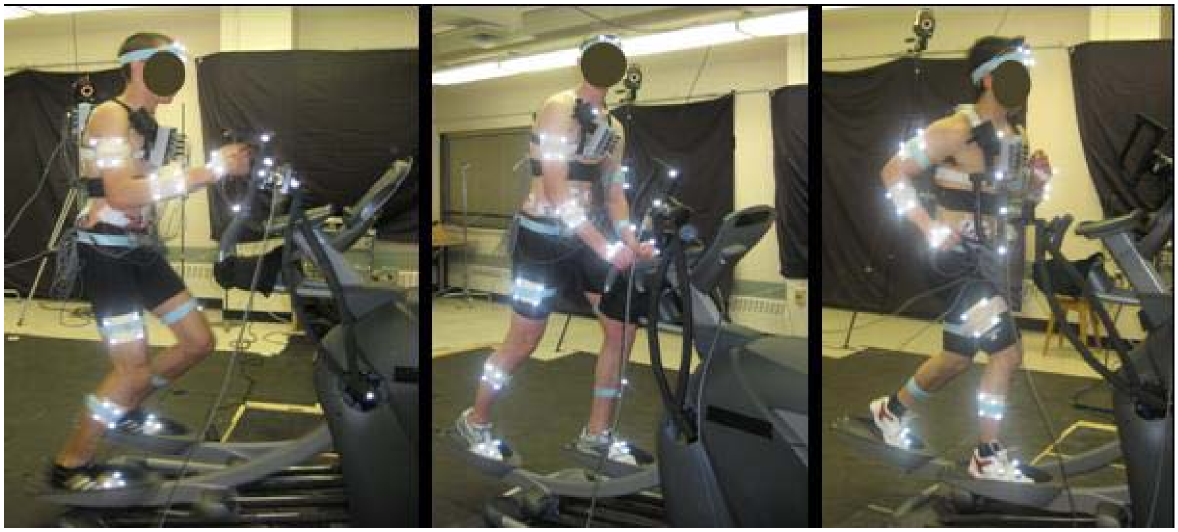


Figure 5-3: The three positions used for testing on the elliptical trainer: “Handles” - On the left, the person is holding onto the oscillating handles of the elliptical. “Bar” - in the second photo, he is holding onto a stationary central bar. “Freehand” - in the third photo, he is not holding on to any part of the apparatus.

Once ready, data collecting moved smoothly from one position to another without stopping in between, although participants were encouraged to alert us if they felt they were getting fatigued and then were allowed to rest until they felt ready to return to exercise. Order of collection was randomized for speed, stride length and hand position. Two collections were obtained for each combination of variables, with approximately 4 cycles of elliptical motion in each.

5.2.7. Kinematics

Motion data were processed using Visual 3D software. 3-dimensional lumbar and hip angles relative to the pelvis were calculated using a Visual 3D algorithm with a Cardan X-Y-Z sequence of rotation (Z up, Y anterior), which uses the method described by Grood and Suntay (1984). Joint angles were filtered with a 6 Hz dual pass Butterworth filter. Signals were screened for abnormalities, processing errors, and abnormal marker movement. Maximum and minimum joint angles were taken from the entire capture time, unless the signal drifted over time due to body position changes (such as neck flexion, which tended to result in increased lumbar flexion), in which case the max/min were extracted from a complete cycle deemed representative of the normal scope of motion. To calculate average joint positions for lumbar and hip flexion angles, trials were clipped to ensure complete cycles of motion (similar to heel strike to heel strike in gait studies).

5.2.8. Kinetics

Prior to being used for kinetic calculations, force and EMG data were down-sampled to 60 Hz, so as to align with the marker position data. A top down rigid linked-segment model was constructed in Visual 3D. In brief, vectors were created which represented the 3 analog forces (x, y and z) and moments for each of the force cubes on the handles of the elliptical trainer. These resultant vectors were transformed into the laboratory coordinate system, then applied to the hand at the mid-hand marker position (approximately mid-way between the 2nd and 5th metacarpal-phalangeal joints). Modeling of the torso included tracking markers at the sternal notch and over C7, in addition to the rigid plate atop T12, to better represent the position of the upper torso. These specific two markers were included in kinetic calculations of lumbar force and moment, but not when calculating lumbar angles. Standard inverse dynamics calculations were carried out to yield resulting forces and moments at the L4, L5 joint, using a Cardan sequence of flexion/extension, side bending, then rotation. Outputs from these analyses, combined with data from the EMG were used to drive an anatomically detailed spine model representing 118 muscle fascicles as well as lumped parameter passive tissues, spanning T12 – L5, S1. This model has been described in detail previously (Cholewicki & McGill, 1996; Kavcic, Grenier, & McGill, 2004a; Moreside et al., 2006). Briefly, using the instantaneous spine position and EMG data, the model calculates individual muscle forces and stiffness, as well as the passive components (due to non-contractile tissues), to provide an internal moment. This is balanced with the external moment from a rigid linked segment model (from Visual 3d), using a least squared error method to calculate a gain factor. This gain is then applied to the internal forces. Total L4, L5 bone-on-bone forces are therefore the sum of the forces due to the external and gained internal moments.

5.2.9. Removing inertial contamination: the effect of the oscillating handles on force cube outputs

It became apparent that the forces and moments being registered by the force cubes in the elliptical handles included a component of inertial force, which was directly attributable to the movement of the handles through space. While this amount was easy to quantify by observing the forces and moments registered in the freehand (FH) condition, it was not possible to simply subtract them from the total forces registered when the handles were being used (H). Visual 3D imports all data in c3d format, which, to the best of my knowledge, cannot be opened and changed. Similarly, the difference in the forces/moments were not necessarily easy to remove as a percentage, nor a pure bias, as they often were more prevalent in one direction than the other (i.e. +ve more so than –ve).

To understand the effect of these inertial components, the data from 2 subjects at 3 varying speeds were analyzed. Subject 9 was our slowest participant, at 40 cps. Subject 11 chose to exercise at 54 and 65 cps in the slow and fast condition respectively, which happened to be the average velocity calculated over all participants. For each participant, forces and moments registered at the right force cube were extracted from two trials: freehand and when the handles were being used.

In that the velocity was standardized between hand position conditions, it was possible to window each trial and overlay the data. To assist with temporal alignment, the position data from the right lateral foot marker was used, as the handles and foot plates move in synchrony. It was not possible to use the handle markers, as they had been removed in the freehand conditions. This marker data allowed us to choose a window of data which represented the same position of the foot, thus the handles in the freehand and handles conditions. The forces and moments registered at the right force cube for these 2 participants (two speeds for #11) are shown in Figure 5-4 and Figure 5-5, respectively.

As can be seen in these figures at slower speeds, the inertial forces and moments are small relative to the forces demonstrated when the person is actually holding the handles. Likely, most of this data can be accepted as being true, and attributable to actual forces applied by the hand to the handle. However, as the velocity increases, so do the inertial forces and moments. There are two things happening here: the faster the handles of the elliptical trainer are being moved through space, the greater the inertial forces and moments will be. Secondly, at the slow speeds, there is greater likelihood that the person exercising will rest their arms on the handles, and actively push/pull the handles, resulting in a greater disparity between the inertial and active forces. As velocity increases, there is less time and for the person to initiate and reverse forces on the handles, thus the arms are basically "going along for the ride", without imparting much additional force.

One of the subsequent problems with this error is that calculation of the lumbar forces and moments are affected by the magnitude of the forces and moments at the handles. Consequently, the faster the elliptical trainer is being used, the greater the error. To gain insight into the effect this error has on the lumbar flexion forces, the inertial force and moment was removed as best as possible as a percentage of the calibration values (initially used to convert volts into newtons). Using data from three trials (velocities of 40, 54 and 65 cps), the graphs representing the 3 forces and moments from each were analyzed and an approximation was made as to how much of the total force/moment could be attributed to the inertial effect. The handles condition was then re-processed in Visual 3D, using these new calibration values, anticipating a change in the value of the forces and moments. These percentage values are shown in Table 5-2. It must be stressed that these are only intended as an approximation and it is recognized that errors will occur by this multiplication, especially if the data alternates in polarity in a different manner between the freehand and handles trials.

Table 5-2: % of force and moment that appears to be attributable to inertial properties, when comparing the freehand and handles conditions at 3 varying speeds. These percentages were subsequently removed from the force cube signals and the data re-processed.

Speed (cps)	Fx	Fy	Fz	Mx	My	Mz
40	10%	25%	25%	5%	10%	___
54	50%	50%	80%	25%	33%	-----
65	66%	33%	90%	25%	33%	20%

In addition, the trials in which the handles were being used were re-processed with all of the forces being removed. Thus, for each of the 3 velocities, the handles trials were processed 3 times:

1. Normal processing, handle forces included
2. Inertial component removed
3. All forces removed

The outcomes (lumbar moments and forces) from these re-processed trials were subsequently input into the spine model, resulting in bone on bone compressive forces. These are shown in Table 5-3.

Table 5-3: Trials representative of 3 different velocities were processed through the rigid link segment model and anatomically detailed spine model three different ways: once will the full forces from the force transducers at the handles intact; “inertial removed” removed a percentage of the handle force and moment based on a visual approximation; “no forces” removed all forces being input from the handles. % difference = the difference between the net calculations and the original “full forces” trial. “rx force” is the reaction force from the rigid linked segment model. A negative shear indicates anterior shear of the thorax on the pelvis.

velocity (cps)			full forces	inertial removed	% difference	no forces	% difference
40	rx force	compression	524	523	-0.19	526	0.38
		shear	-96	-91	-5.21	-78	-18.75
	muscle force	compression	1370.8	1370.8	0.00	1462	6.65
		shear	336.1	336.1	0.00	358.5	6.66
	Total bone on bone	compression	1894.8	1893.8	-0.05	1988	4.92
		shear	240.1	245.1	2.08	280.5	16.83
54	rx force	compression	427	420	-1.64	418	-2.11
		shear	-101	-101.6	0.59	-99	-1.98
	muscle force	compression	1387	1387	0.00	1387	0.00
		shear	334	334	0.00	334	0.00
	Total bone on bone	compression	1814	1807	-0.39	1805	-0.50
		shear	233	232.4	-0.26	235	0.86
65	rx force	compression	430	410	-4.65	407	-5.35
		shear	-105	-110	4.76	-105.9	0.86
	muscle force	compression	1540	1232	-20.00	1386	-10.00
		shear	351	281	-19.94	316	-9.97
	Total bone on bone	compression	1970	1642	-16.65	1793	-8.98
		shear	246	171	-30.49	210.1	-14.59

Based on these sensitivity calculations, it appears that at low and normal velocities, when there is the time and tendency to exert a force on the handles, and inertial forces are low, the “full forces” outcomes are not significantly different from those calculated with the inertial forces removed (less than 5% difference). However, as velocity increases, so do the inertial effect and error: at 65 cps, the difference between the “inertial removed” calculations and the original was in the order of 17 – 30%.

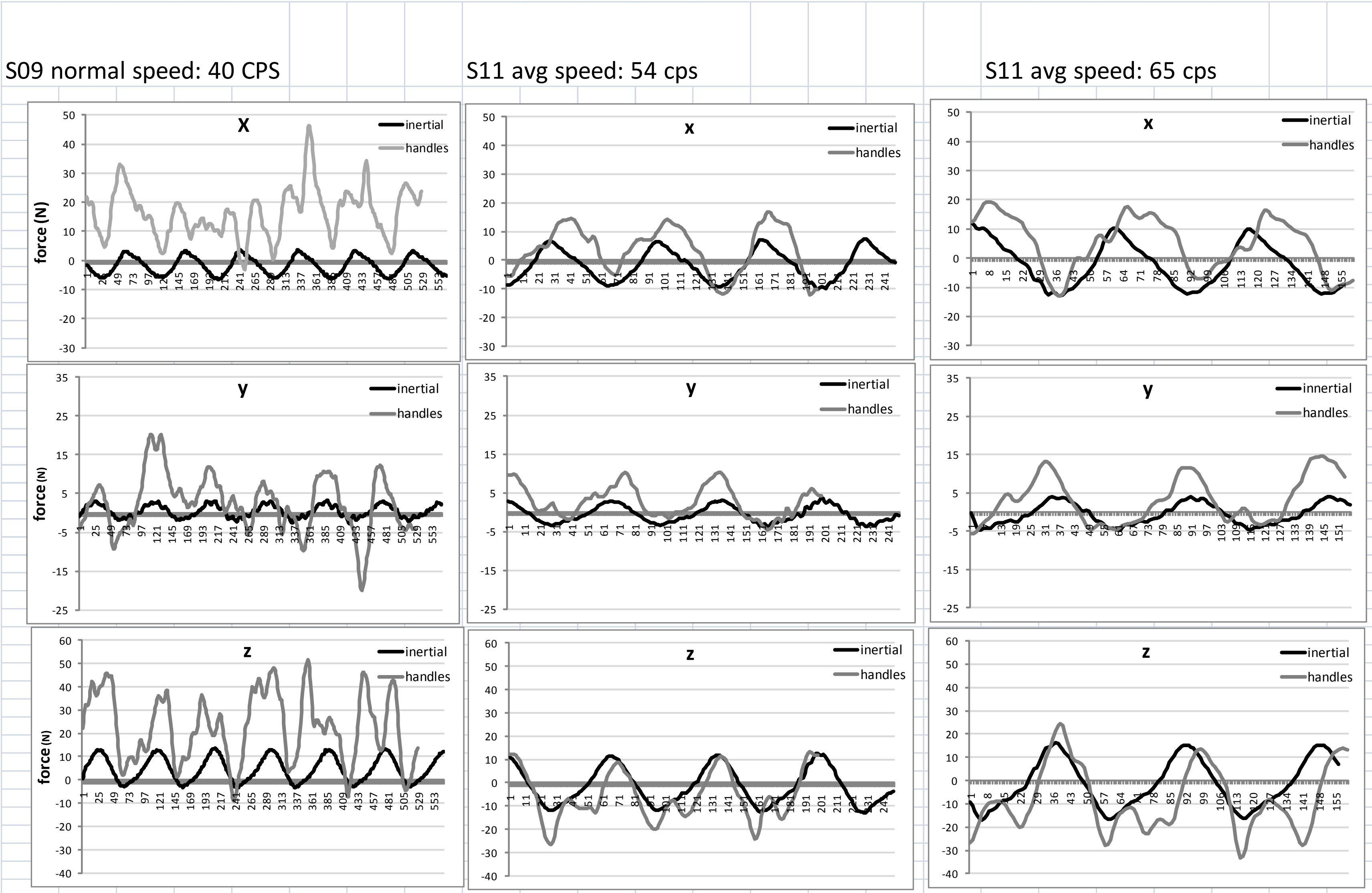


Figure 5-4: Force data registered at the force cube on the right handle of the elliptical trainer. The 3 columns represent 3 different velocities, ranging from 40 – 65 cycles per second. The dark line represents the forces due to the inertial effect of the handle swinging back and forth, when the participant when not holding onto the handles. The lighter line represents the forces when they person was holding the handles. In each graph, data is being compared between two trials with the same velocity; data was windowed and aligned to approximate the same time of the cycle.

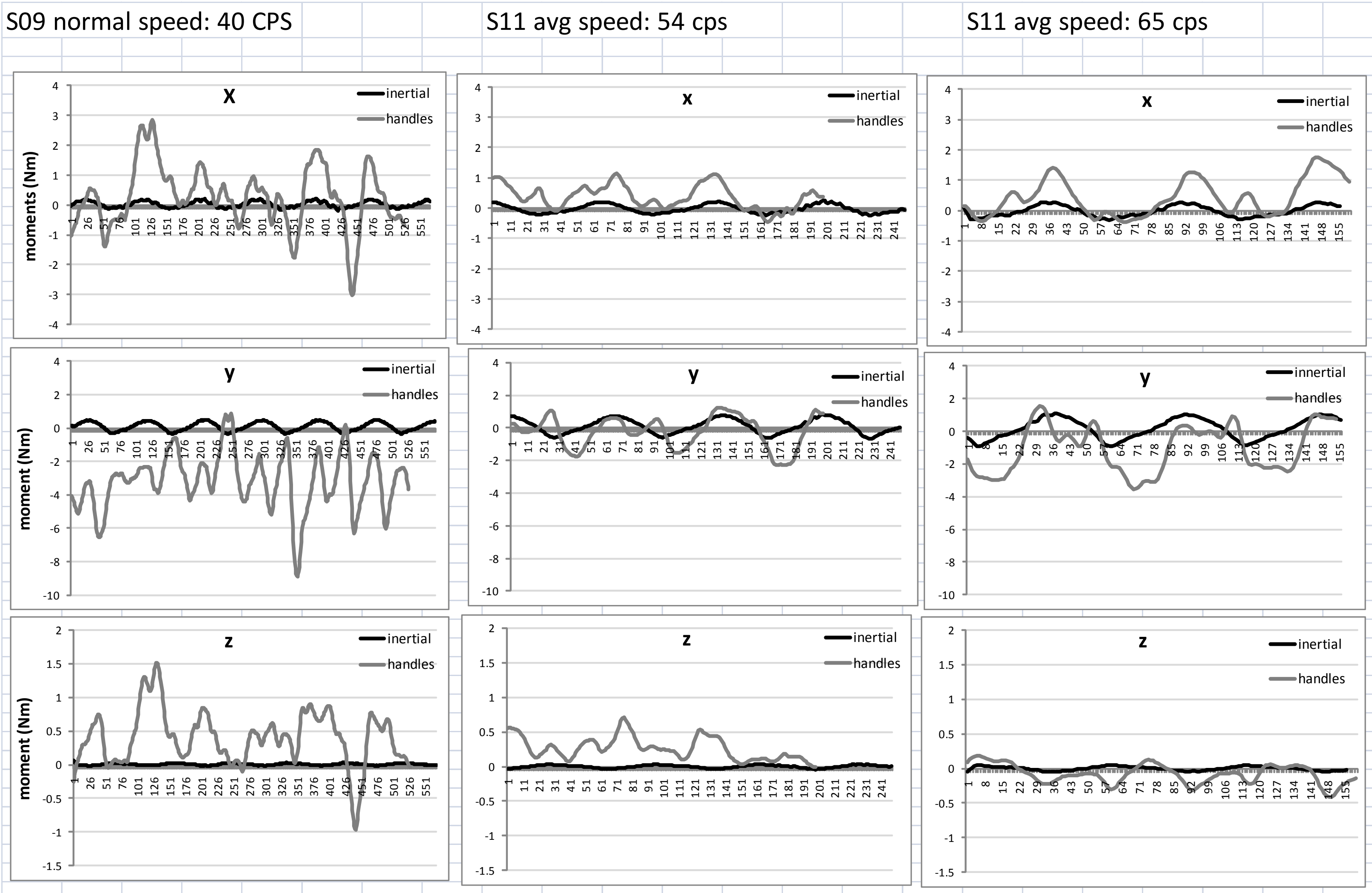


Figure 5-5: Moment data registered at the force cube on the right handle of the elliptical trainer. The 3 columns represent 3 different velocities, ranging from 40 – 65 cycles per second. The dark line represents the moments due to the inertial effect of the handle swinging back and forth, when the participant when not holding onto the handles. The lighter line represents the moments when they person was holding the handles. In each graph, data is being compared between two trials with the same velocity; data was windowed and aligned to approximate the same time of the cycle.

5.2.10. *Statistical analysis*

Independent samples t-tests were used to compare body height, mass, average hip rotation and extension measurements between the two groups. A repeated measures ANOVA with LHM/EHM as the between subject factor was conducted to compare the angle of the pelvis between groups in two standing trials: that in the standing calibration trial, and the position of the pelvis averaged over the first 10 frames of data (prior to initiation of movement) in the "twist and reach" trial. Active lumbar ROM in the 3 anatomical planes was compared between the two groups using independent samples t-tests with Bonferroni adjustments. During the active movement trials, repeated measures ANOVAs were conducted on lumbar and hip angles, with LHM/EHM as a between subject factor. This same test was used on the elliptical trials to determine differences in average hip and back angles, and lumbar forces between the two groups. For hip joint calculations on the elliptical trainer, right/left symmetry was assumed and calculations were performed on the right leg.

Paired t-tests were conducted, comparing the right and left sides of the body, on peak EMG levels when the participants were using the elliptical trainer. In that no significant differences were found, the sides were collapsed and independent samples t-tests were used to determine differences between the LHM and EHM groups for each muscle.

5.3. Results

There were no significant differences between the groups for height and mass (p = 0.988 and 0.243, respectively). Despite the non-significance, the mean mass was higher in the

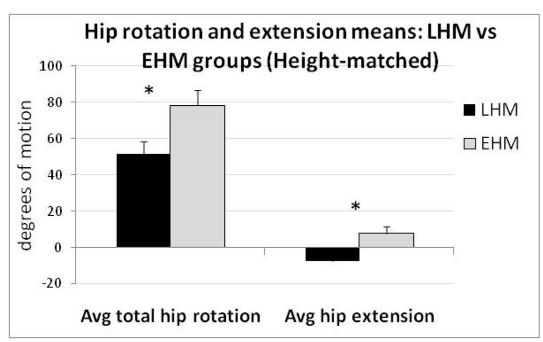


Figure 5-6: Comparison of average hip extension and total rotation (IR + ER) between the LHM and EHM groups. (n = 10 per group)

LHM group: 84.6(15) kg compared to 78.0(7) kg in the EHM. Mean height was identical between groups: 180.1(7) cm. Hip mobility was significantly different between the two groups in both extension and rotation, with p – values of < 0.001. Figure 5-6 shows comparison of hip mobility, averaged across participants in each group (See Appendix 5-1 for individual hip ROM, means and SDs).

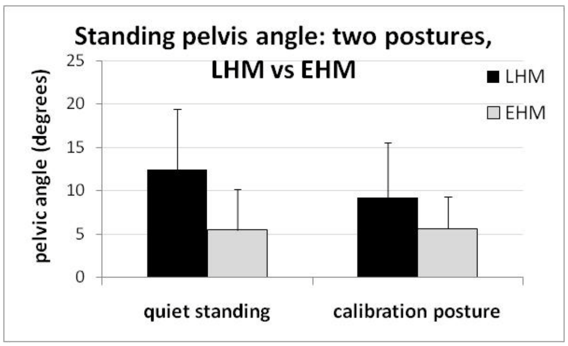


Figure 5-7: Pelvic angle as calculated as the angle between the 2 ASISs and 2 PSISs. (n = 10 per group)

Pelvis angle in standing was significantly different between the two groups when observed in the calibration posture and quiet standing (p = 0.037) (Figure 5-7). The LHM group stood with an anterior tilt of 12.4(7)° and 9.2(6)° in quiet standing and calibration pose, respectively, compared to 5.5(5)° and 5.6(4)° in the EHM group. These angles for the EHM group are similar to those described by Schache et al (2003): 5.4° in a group of 22 males (mean age 35(7) years), using the same bony landmarks to calculate standing pelvic angle. The comparison between relaxed standing and the calibration pose was conducted as it became apparent throughout this, and associated studies, that the calibration posture may not accurately represent a person's normal standing position. The calibration posture requires each participant to stand upright with the arms abducted approximately 90°. The question arose as to whether this more formal posture would induce a different angle in the pelvis (thus also low back and hips) than that adopted in relaxed standing, when the participants were not thinking about posture. The markers used for calculating this angle were the 2 ASISs and 2 PSISs. As can be seen in Figure 5-7, although there was no significant difference between the two postures, the LHM group did tend to anteriorly tip their pelvis more in relaxed standing; thus in the calibration pose, they would "brace" the system by tucking their pelvis under (posterior rotation). This effect was not observed in the EHM group, and the variance is high, but it does lend credence to the theory that caution should be used when choosing hip and lumbar angles extracted from the calibration posture to define 0° at these joints.

There were no significant differences in the available active lumbar ROM between the groups (*p*- values ranged from 0.141 to 0.390). As shown in Figure 5-8, the tendency was for the LHM group to demonstrate less ROM in all directions. Average ROM for both groups, in each direction, are specified in Table 5-4.

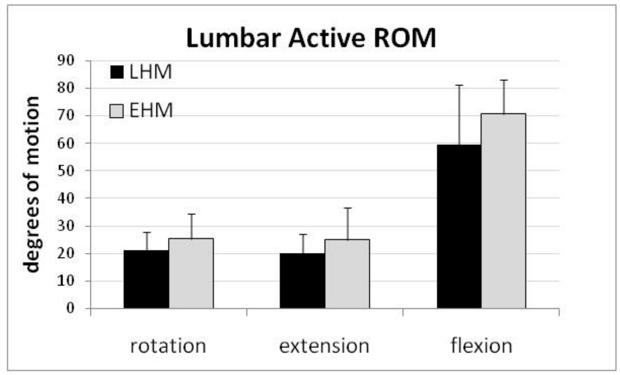


Figure 5-8: Average lumbar ROM; rotation is the sum of right and left combined. (n = 10 per group)

Table 5-4: Average active lumbar ROM comparing the LHM and EHM groups. Total rotation is the sum of rotation to both sides.

	Total rotation	Extension	Flexion
LHM	21.2(7)	20.1(7)	59.6(22)
EHM	25.3(9)	25.0(12)	70.6(12)

Right hip and lumbar angles were calculated during 4 specific movement trials: walking, lunging (non-dominant leg in front), active hip extension (dominant leg), and "twist and reach" (described earlier in *section 5-2-6*). For the hip outcomes, right/left symmetry was assumed for the twist and walking trials, and calculations were performed on the right leg. For the lunge and hip extension, the dominant leg outcomes are presented. Graphs depicting average hip and back motion during these trials appear in Figure 5-9. There were no significant findings with regard to differences between the LHM and EHM groups. Again, there was a trend seen in walking, lunging and twisting that the group with LHM moved more

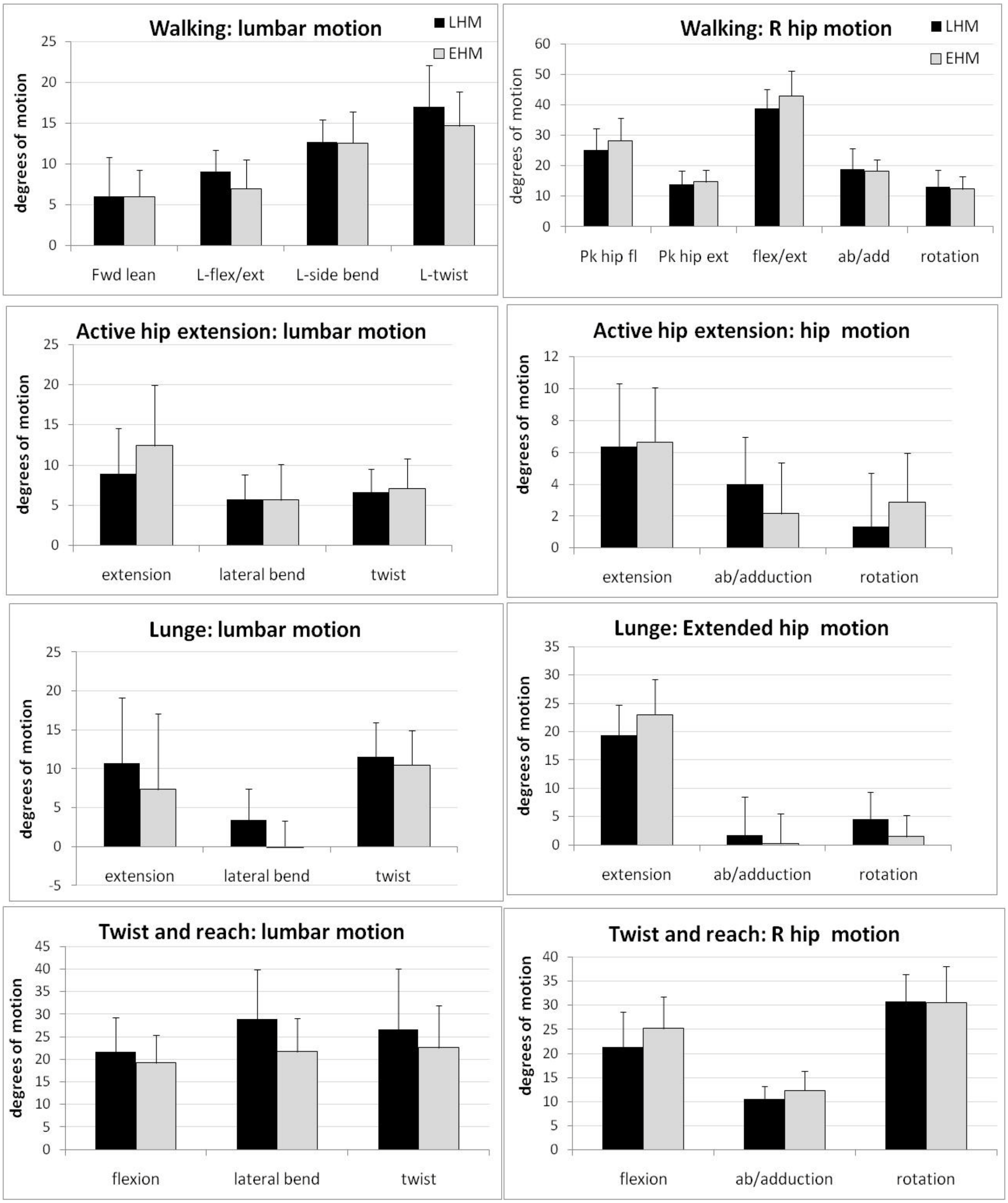


Figure 5-9: Lumbar spine and hip motion during the 4 non-elliptical movement trials. In the “twist and reach” trials, lateral bend and twist are the combined totals of twisting to the right and left. Angle measurements in the hip extension and lunge trials were captured at the time of peak hip extension. (n = 10 per group)

in their back, less in their hips than the EHM group. This trend was reversed, however, when asked to demonstrate active hip extension.

On the elliptical trainer, the LHM group had significantly more peak lumbar flexion, minimal lumbar flexion, and average flexion angle than the EHM group (p = 0.019, 0.016 and 0.015, respectively) (Figure 5-10, Table 5-5). However, there was no significant difference between groups in the range of flexion/extension, lateral bend or twist in the lumbar spine (p = 0.529, 0.380 and 0.972, respectively). Basically, the LHM group adopted a more flexed posture, but the total range of motion covered with each cycle did not differ between groups.

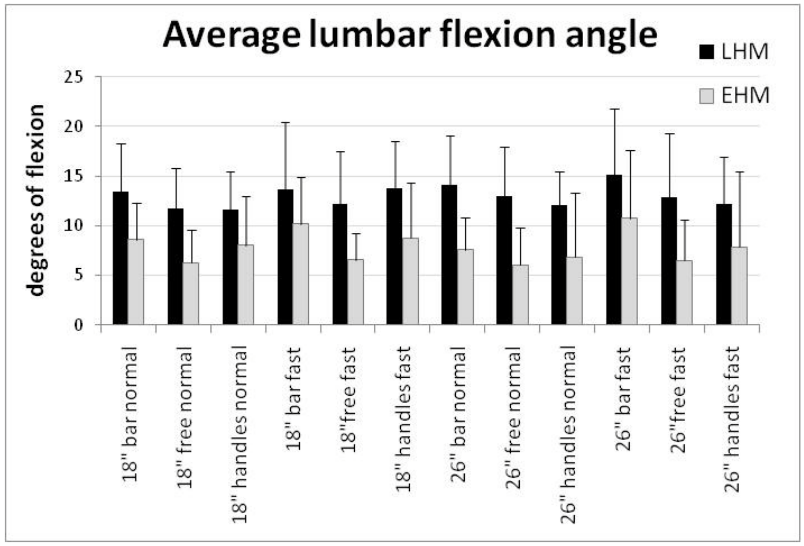


Figure 5-10: Average lumbar flexion angle on the elliptical trainer, demonstrating the significant difference between the LHM and EHM groups across all conditions (p = 0.015). (n = 9 per group)

Table 5-5: Average lumbar motion on the elliptical trainer, averaged over 9 participants in each group, and collapsed over the different hand/stride/velocity conditions. * indicates a significant difference between the LHM and EHM groups ($p < 0.05$)

	LHM	EHM	Mean difference
Forward lean *	12.9 (5)	7.8 (5)	5.1
Peak flexion *	16.7 (6)	11.4 (6)	5.3
Minimum flexion *	8.9 (5)	4.0 (5)	4.9
Total sagittal	7.8 (2)	7.5 (2)	0.3
Total frontal	9.5 (2)	7.8 (2)	1.7
Total twist	18.5 (5)	18.5 (5)	0

Also on the elliptical trainer, the LHM group used significantly less hip flexion/extension, demonstrated a smaller average hip flexion angle and peak hip flexion (p =

0.048, 0.010 and 0.009, respectively). Peak hip extension, total ab/adduction and total rotation showed no significant differences between the groups (p = 0.124, 0.427 and 0.845, respectively) (Table 5-6).

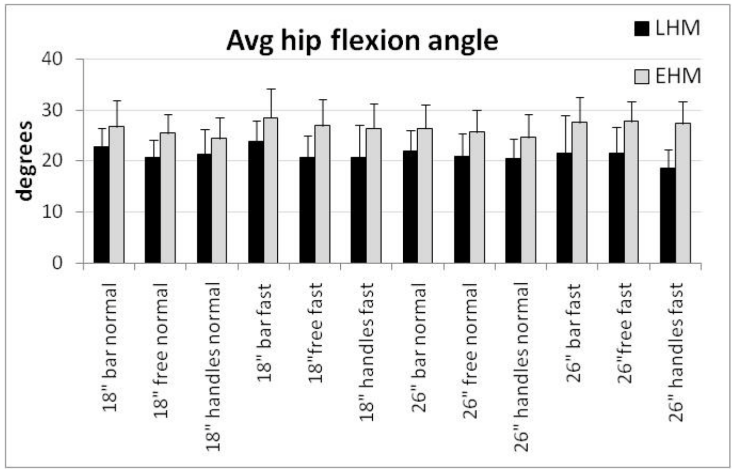


Figure 5-11: Average hip flexion angle on the elliptical trainer. The EHM group adopted a posture with significantly more hip flexion (p = 0.0.010). (n = 9 per group)

Table 5-6: Average hip angles on the elliptical trainer, averaged over 9 participants in each group, and collapsed over all the stride/velocity/handles conditions.
* indicates a significant difference between the LHM and EHM groups ($p < 0.05$)

	LHM	EHM	Mean difference
Avg flexion angle *	21.3 (6)	26.5 (6)	-5.2
Peak flexion *	44.2 (2)	50.3 (2)	-6.2
Minimum flexion	-3.0 (5)	0.2 (5)	-3.3
Total sagittal *	47.2 (5)	50.6 (5)	-3.4
Total frontal	17.1 (6)	15.6 (6)	1.5
Total twist	11.5 (4)	11.7 (4)	-0.2

Muscle activation levels on the elliptical were compared between groups using the 26”/handles/fast condition, as previous analysis had demonstrated that it resulted in the largest amount of activity for most muscles (see Chapter 9). There were no significant differences between the groups, although the EHM group did

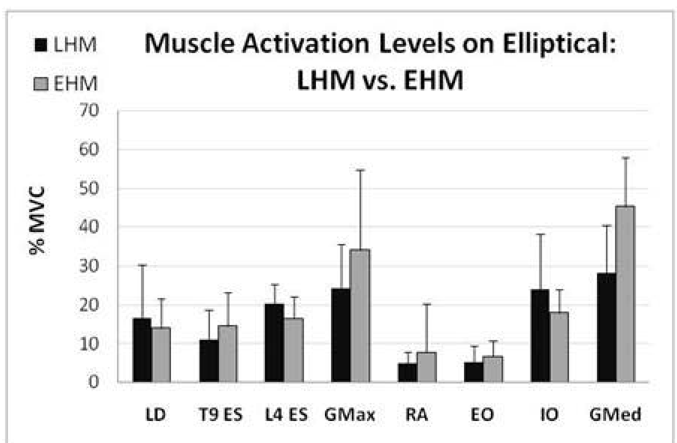


Figure 5-12: Average muscle activation levels, comparing the LHM and EHM groups. The 26”/handles/fast condition was used for this data set. (n = 9 per group)

tend to recruit more gluteal activity, especially GMed, with a mean of 45.3% MVC compared to 28.2% in the LHM group (**Error! Reference source not found.**). Using paired t-tests, this resulted in a p-value of 0.025. However, the Bonferroni adjustment over 8 muscles yields a required p-value of 0.006 for statistical significance. The statistical power for this calculation was 0.36; however, the effect size (Cohen's d) was 1.4, indicating a large effect of hip mobility on GMed activation. The increased lumbar flexion angle in the LHM group is also reflected in a non-significant increase in the L4 ES muscle group.

Calculating lumbar forces and moments on the elliptical trainer presented a unique set of methodological problems, as described in section *5.2.9.* Briefly, the forces and moments registered at the force cubes included a certain amount of error due to the inertial effect of the handles moving through space. This error increased with the velocity of the activity. Consequently, kinetics will only be presented from the normal speed elliptical trials when the handles were being used. Similarly, the central bar that the participants held on to in the "bar" conditions, was not instrumented with a force transducer, that it is not possible to calculate kinetics in these trials. In association with the increased spine flexion demonstrated by the LHM group, the anterior/posterior shear reaction forces at L4, L5 tended to be higher in the LHM group (p = 0.076) (Figure 5-13) with the LHM group averaging -1.3(6) N/kg, compared to -9.1(6) N/kg in the EHM group, when

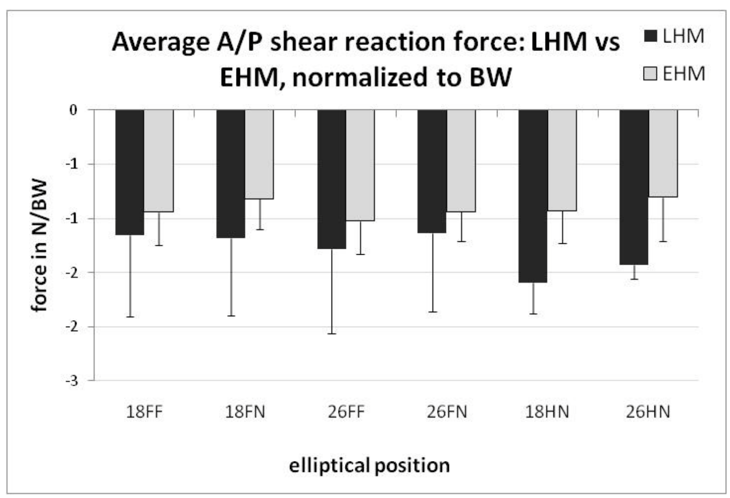


Figure 5-13: Average anterior/posterior shear reaction force on the elliptical trainer, normalized to body weight. Negative shear represents anterior shear of the thorax relative to the pelvis. (n = 9 per group)

collapsed over the various elliptical conditions. The statistical power in this comparison was only 0.43, likely due to the large variability that exists, despite the consistently greater shear force demonstrated by the group with limited hip mobility. Similarly, the Cohen's d effect size was 0.45, indicating a medium effect of hip mobility on A/P reaction force shear. Normalization to body weight was conducted to account for the 6.6 kg difference in average mass between the groups. Vertical compression (normalized to BW) between groups was almost identical: mean compression was 5.077 N/kg for the LHM group and 5.096 N/kg for the EHM (p = 0.842).

To further investigate the effect that the increased spine angles and muscle activation patterns have on total spine compression and shear, data were input into an anatomically detailed spine model, described in section *5.2.8.* Four participants were chosen: two from the

	LHM			EHM		
subject #	**#11**	**#34**	avg	**#26**	**#40**	avg
mass	79.9	95		79.5	94	
height	181	184		182	185	
velocity	53	50		48	49	
18HN						
Lumbar angle	10	13		6	10	
thorax angle	9	18		9	12	
compression	1858	1660	1759	1528	1383	1455
A/P shear	227	192	210	201	166	184
18FN						
Lumbar angle	9.7	10		7	2	
thorax angle	9.6	14		7	6.7	
compression	1767	1756	1762	1304	1568	1436
A/P shear	193	227	210	138	124	131

Table 5-7: Bone on bone compression and shear forces (N) for two participants from each group, matched for height and mass. Two conditions were analyzed: 18HN = 18" stride, using the handles, normal velocity. 18FN = 18" stride, normal velocity, but the hands are not holding on. Positive shear forces represent a posterior shear of the thorax relative to the pelvis.

EHM group and two from the LHM group, who demonstrated similar velocities, masses and heights. Two trials were analyzed, both being an 18” stride, normal speed: one with handles and one freehand. As can be seen in Table 5-7 and Figure 5-14, when the forces from the rigid linked segment model and those attributed to the muscular and passive tissues are combined, the LHM group demonstrated higher bone on bone compressive and shear forces than the EHM group. These force outputs are dominated by the muscular/soft tissue components by a ratio of approximately 3 to 1 (soft tissue vs. those from the rigid linked segment model). Further analyses of individual muscle forces from the spine model demonstrates that the increased compression force is associated with increased force production in the back muscles: the ES group, QL, Mult, as well as psoas in the LHM hip group (Table 5-8).

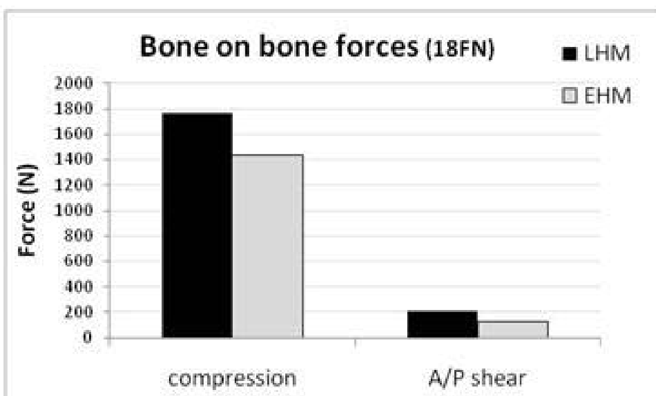


Figure 5-14: Total compression and shear forces, averaged over 2 participants in each of the LHM and EHM groups. This represents data from the 18” stride/freehand/normal velocity trial.

Table 5-8: Muscle forces (N) produced by specific muscle groups unilaterally, as calculated by an anatomically detailed spine model. Model inputs included lumbar angle and EMG data averaged over one complete elliptical cycle. Model outputs were averaged over 2 participants in each group, with right and left sides collapsed.

		RA	**EO**	**IO**	**ES**	**QL**	**LD**	**Mult**	**Psoas**
18HN	**LHM**	33.4	26.3	70.7	356.2	73.1	63.7	62.4	233.8
	EHM	25.9	33.0	78.6	299.1	53.3	69.4	47.4	215.2
18FN	**LHM**	34.3	35.3	98.7	358.7	73.5	63.9	64.7	273
	EHM	26.5	35.8	78.0	278.2	53.9	66.1	47.9	200.9

5.4. Discussion

Studying two groups of young men with significantly different hip mobility gives us insight into how these differences are manifested in functional movements, and the extent to which they affect the lumbar spine. To start with, men with LHM stand differently: their average anterior pelvic tilt that was approximately 7° greater in relaxed standing than those in the EHM group. This would tend to result in an increased lumbar lordosis and subsequent greater compression in the posterior structures of the lower lumbar segments. Perhaps to compensate for this, these same participants increased the flexion of their lumbar spine when exercising on the elliptical trainer. While possibly reducing the tendency for facet joint compression, the penalty for this posture is increased activity in the back extensor muscles, resulting in substantially greater compressive force and posterior shear of the thorax relative to the pelvis. The resulting decrease in facet joint compression will tend to increase the percentage of being load transmitted through the intervertebral disc. This, combined with increased available lumbar rotation (due to decreased facet joint abutment) may result in abnormally high amounts of lumbar rotation being born by the disc; a structure that is not designed to withstand repetitive flexion and rotation. Thus, having hips with limited mobility ultimately resulted in more of a flexed lumbar spine, and a subsequent increase in the lumbar compressive and shear loads. Patients who are intolerant of lumbar flexion, rotation, and/or increased lumbar compression should be aware of how hip mobility affects these parameters.

Available active lumbar ROM was not different between the groups (**Error! Reference source not found.**) although there was a tendency for the EHM group to have more mobility. If one considers, however, that the pelvis is in a more anteriorly tilted position prior to the initiation of movement in the group with LHM, it follows that the lumbar spine would likely also be more lordotic (Dunk et al., 2009; R. Y. W. Lee & Wong, 2002; Perry, 1992). This would tend to limit the amount of available rotation and extension (Öhlen, Wredmark, & Spangford, 1989) during active motion, due to facet joint encroachment. Thus, to conclude from Figure 5-8 that the group with limited mobility tended to have less extension and rotation might be erroneous as it does not standardize for pelvis and lumbar spine position prior to motion. However, the non-significant larger amount of flexion in the EHM group tends to indicate that those who have EHM in the soft tissues of the hip are also more flexible in the posterior lumbar support structures. It would be of interest to look further into those with hip mobility differences, and determine if the same trend exists for joint mobility throughout the body.

Although there were no significant differences between the two groups in terms of hip/spine motion during the non-elliptical trials, (Figure 5-9) the trends do tend to support the hypothesis that the LHM group utilized more spine motion compared to the EHM. During walking, lunging, and twisting/reaching, the LHM group consistently demonstrated more spine flexion/extension and rotation than the EHM group. Similarly, the EHM group used more hip flexion/extension during these motions. The bar graphs of Figure 5-9, however, belittle the importance of the temporal qualities of motion, and the large variability between subjects. Figure 5-15 shows hip and back data from three separate participants when they were asked to extend their hip (+ve direction indicates more extension for both the hip and back). In the first graph, initial spine flexion is associated with relative hip extension as the person braces for un-weighting the leg. This person likely rotated their pelvis posteriorly prior to lifting. Although he is successful at extending the hip past the initial starting position, most of the backward motion of the leg is accomplished via lumbar extension. The second graph demonstrates lumbar and hip flexion prior to un-weighting, assumedly flexing the upper body and pelvis forwards. This person is unable to demonstrate any further hip

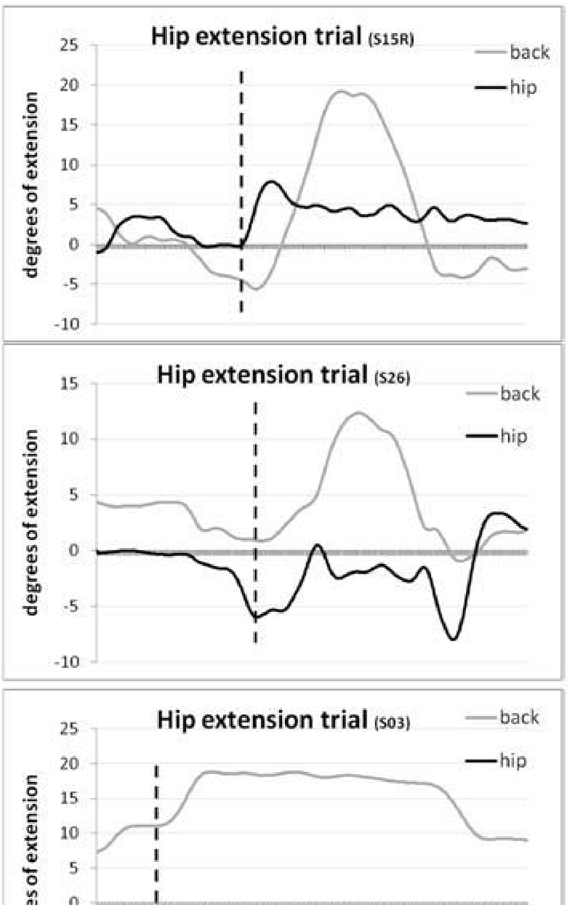


Figure 5-15: Examples of variability of hip/spine movement patterns elicited when 3 different participants were asked to "extend your hip". Dashed line represents unweighting of the leg. A +ve direction indicates extension of both the hip and back.

extension than in his initial standing position, until he returns to relaxed standing at the end of the trial, as he uses his lumbar spine to accomplish most of the leg motion. The third graph was a participant we specifically asked to not move his pelvis/spine while extending the leg (this was done in addition the regular hip extension trial). As you can see, what he perceived to be hip extension was entirely accomplished by lumbar spine motion; in fact, his hip flexed relative to the pelvis, despite the backward motion of the leg in space. These graphs highlight the different motor patterns adopted by 3 people in what would at first appear to be a simple command, and the difficulty in looking for trends in voluntary movements. In addition, the lack of true hip extension in this maneuver should be considered when using active hip extension to determine full available range of motion for normalizing hip extension.

The elliptical trainer provided a medium which allowed standardization of stride length and hand position. In that the trials were captured while the participant was already in motion, there were not the same problems with "setting the spine", as we saw in the other motion trials. The LHM group adopted a posture of increased lumbar spine flexion, and decreased hip flexion (Figure 5-10, Figure 5-11). This decrease in mean hip flexion angle may be a direct result of posterior pelvis rotation associated with lumbar flexion, thus causing a decrease in hip flexion angle relative to the pelvis. Interestingly, the mean difference in lumbar spine angle between groups is 5.4°, which, if it is due to posterior pelvic rotation, may bring the lumbo-pelvic angle back to one that is more similar to the EHM group in the initial standing posture (Figure 5-7). The significant finding of more sagittal hip motion in the EHM group is the first instance in this study to demonstrate this group actually utilizing their greater available hip ROM during functional movements. Previous literature has also documented weak correlations between hip extension measured in the MTT and that demonstrated during gait and running (r = 0.41 by Lee et al, (1997) and no significant correlation as in Schache et al., (2000)). The standardization of stride length and hand position in this study may have reduced the confounding effect of other joint compensatory movements during an activity that is relatively similar to gait. While it was anticipated that the elliptical would encourage full extension of the hip joint, it appears that less extension is required than that found in normal walking. In a previous study, analysis of 40 young adult males exercising on the elliptical trainer found that the most hip extension demonstrated in any of the stride/handles/speed conditions was an average of 7°, compared to 13° demonstrated by the same group in normal walking (Chapter 9-4). Consequently, these young men did not reach maximum hip extension; nearing the end of hip range might have resulted in different lumbar spine compensatory mechanisms.

While not attaining statistical significance, it is interesting to note the increased activation levels of the gluteal muscles in the EHM group, when using the elliptical trainer. Potentially, those with decreased flexibility in the hip rotators and flexors would be able to access the tension in the passive tissues to maintain hip joint stability, whereas the looser group would rely more on active muscle contraction. This is in keeping with the work of Gleim, Stachenfeld and Nicholas (1990) who demonstrated that men and women with increased trunk and lower limb flexibility (based on a total of 11 flexibility tests) were significantly less efficient metabolically during walking and treadmill running than their less flexible counterparts. Increased L4ES and IO activity in the LHM group, despite its non-significance, infers an increased co-contraction of these trunk stabilizers, which ultimately resulted in increased lumbar compressive loads. Although only calculated on a few participants at this point in time, this was indeed the case when bone on bone compressive and shear forces were compared between the groups. The increased lumbar forward lean would result in greater shear loads, requiring larger amounts of muscle activity to resist the shear, and ultimately greater spine compressive loads.

Limitations exist, in that this study only looked at LHM/EHM differences in a group of young male adults. This was done to limit the variability that might occur between sexes as well as the possibility of changes in ligamentous laxity over the menstrual cycle which has been described in females (Chandrashekar, Slauterbeck, & Hashemi, 2005; Chandrashekar, Mansouri, Slauterbeck, & Hashemi, 2006; J. Slauterbeck, Clevenger, Lundberg, & Burchfield, 1999; J. R. Slauterbeck et al., 2002; Wojtys, Huston, Boynton, Spindler, & Lindenfeld, 2002). A larger sample population might have lent more statistical power to the analyses. There were many non-significant trends depicted in this study. Recruiting participants who fit into extreme groups, while maximizing the difference between groups, also limits the number of participants who qualify for participation. Despite the height matching of the two groups, the body types may have differed. The LHM group participants who volunteered for this study tended to be highly athletic and were heavier, on average. Specific movement patterns adopted may have been the result of engrained sports-related activities. Finally, the large variability that exists when studying human movement often precludes statistical significance, but can highlight trends that are important to a clinical population.

5.5. Summary

Limited hip joint mobility affects the lumbar spine. In relaxed standing, those with LHM stood with greater anterior tilt of the pelvis. During dynamic exercise on the elliptical trainer, they exhibited greater amounts of lumbar flexion than the EHM group, adopting a posture of increased forward lean. This resulted in an increased compressive load on the lumbar spine due to the higher back muscle forces required to counterbalance the flexion. Consequently, the elliptical trainer may not be the exercise modality of choice for those who have limited hip mobility combined with lumbar flexion and/or compression intolerance. Conversely, those with limited hip motion who prefer lumbar flexion (such as an older, spondylitic group) may find it a safe exercise medium, which challenges the hip stabilizing muscles, yet requires less hip extension than walking. In activities other than the elliptical trainer, it remains difficult to identify significantly different movement patterns between the LHM and EHM groups, as individual variability in movement patterns is substantial.

5.6. Appendix

<u>Appendix 5-1:</u> *Participant anthropometrics in the EHM, or loose (L) group and LHM, or tight (T) groups. Extension measurements are relative to the horizontal.*

Subject #	group	height(cm)	mass(kg)	R rot	L rot	Avg rot	R ext	L ext	Avg ext
33	L	169	69	83	86	85	-6	-6	-6
19	L	172	76	82	77	80	-4	-4	-4
20	L	175	77	79	71	75	-11	-8	-10
24	L	179	78	91	99	95	-10	-9	-10
8	L	180	73	78	82	80	-4	-6	-5
23	L	182	82	65	77	71	-4	-5	-5
26	L	182	79	76	82	79	-15	-16	-16
41	L	183	70	78	74	76	-5	-4	-5
40	L	185	94	64	64	64	-4	-8	-6
27	L	195	82	79	81	80	-14	-12	-13
average		**180.1**	**78.0**	**78**	**79**	**78**	**-8**	**-8**	**-8**
st dev		7.4	7.1	8.0	9.4	8.2	4.4	3.8	4.0
21	T	170	54	52	55	54	5	5	5
9	T	172	102	42	45	44	14	12	13
13	T	175	75	39	42	41	8	12	10
38	T	178	101	49	50	50	8	3	6
29	T	179	79	56	53	55	6	4	5
11	T	181	80	46	71	59	10	12	11
5	T	183	95	40	50	45	5	5	5
34	T	184	95	47	53	50	6	9	8
37	T	186	93	57	64	61	5	16	11
15	T	194	72	53	64	59	4	3	4
average		**180.1**	**84.6**	**48**	**55**	**51**	**7**	**8**	**8**
st dev		7.1	15.3	6.4	9.1	6.9	3.0	4.7	3.3

6. The effect of hip flexibility and core stability on lumbar spine motion: A trial to enhance hip mobility and possible function

6.1. Introduction

Rehabilitation and fitness workers often focus on improving hip flexibility and improving core strength, with the assumption that this will assist injury prevention. But it is also understood that movement patterns are the result of a lifetime of experience: both physical and emotional. It is thought that patterns of movement develop that are energy efficient, relying on passive structures for energy storage, musculo-tendinous structures for generation and control of movement, and neurological control to coordinate smooth movement. The question arises as to whether these improvements in flexibility and/or strength will transfer over to function. Specifically, if a person presents with limited hip mobility, is there any evidence that improvements in hip range of motion (ROM) will be utilized during functional activities? The literature indicates that hip extension measurements obtained passively do not reflect those utilized during dynamic activity (L. W. Lee et al., 1997; Schache et al., 2000) but there is little information as to whether improvements in hip extension and/or rotation over time will be reflected in functional activity. For a new motor pattern to become spontaneous, old patterns of movement must be overcome (Caillou, Nourrit, Deschamps, Lauriot, & Delignieres, 2002; Delignières et al., 1998; Nourrit, Delignieres, Caillou, Deschamps, & Lauriot, 2003; Vereijken, van Emmerik, Bonnefoy, Beek, & Newell, 1997) thus it may not be enough simply to improve hip mobility or spine stabilization, without specifically focusing on preferred movement patterns.

Limited hip mobility is known to cause secondary positional changes in the pelvis and lumbar spine (Kerrigan et al., 2003; L. W. Lee et al., 1997; Martin et al., 1997; Murray, 1967; Perry, 1992; Thurston, 1985), although most previous research has focused on either the aging arthritic hip or those with increased hip flexor tightness due to neurological impairments. Clinically, there also exist young adults who fall into neither pathological group, yet demonstrate limited hip range of motion (ROM) into extension and rotation, either due to congenital factors or possibly resulting from local muscle hypertrophy. Clinical interventions aimed at stretching soft tissues that affect the hip joint have been shown to result in significant increases in muscle length and joint mobility (Bandy et al., 1994; Bandy et al., 1997; Feland et al., 2001; Madding et al., 1987; Prentice, 1983; Roberts & Wilson, 1993; Sady et al., 1982), although the majority of this research has specifically addressed hamstring tightness, which would limit neither extension or rotation. Improving core trunk muscle strength and awareness may also assist in giving the body a stable base upon which to optimize hip motion. Changes in spine posture and gait subsequent to a 12 week rehabilitation regime has been documented (Scannell & McGill, 2003), but limited information exists as to whether these localized muscle tension changes can transfer over to the movement patterns utilized in other activities, or whether changes in hip mobility will result in spine motion changes.

The purpose of this study was to analyze the effect of a 6 week specific hip stretching/spine stabilizing protocol on lumbar spine motion during functional activities, in a group of young males with limited hip mobility. The aims of the 4 treatment protocols were as follows:

- Group 1: increase passive hip extension and rotation

- Group 2: increase passive hip extension and rotation, while improving their motor control abilities by disassociating trunk rotation and flexion/extension from hip motion
- Group 3: improve trunk muscle strength and endurance, and improve hip/back disassociation
- Group 4: control group received no treatment protocol during the 6 weeks

The hypotheses being tested were:

1. Six weeks of stretching will result in a significant improvement in passive hip extension and rotation.
2. Six weeks of strengthening exercises will result in a significant improvement in core strength (group 3), and documentable progression of motor control exercises, based on meeting progression criteria (group 2).
3. Increased passive hip mobility will result in increased hip extension and rotation utilized during functional movement tests.
4. Improved core strength and motor control will result in a decrease of lumbar rotation and flexion/extension during functional movement tests.

6.2. Methods

6.2.1. Participant selection

Participants were recruited from the university population and surrounding area via posters and word of mouth. All subjects were healthy without current hip or back pain or past pathology in these regions. Participants completed a written informed consent document approved by the University of Waterloo Office for Research Ethics.

Previous research in this laboratory analyzed hip extension and prone lying hip rotation in a group of 77 males, between the ages of 19 and 30 to determine normative and percentile data for this age group. Results are shown in Table 6-1. In that the purpose was to study a group of young males with limited hip mobility and observe changes that ensue with a

Table 6-1: Percentile data of hip extension and rotation from a young adult male population. 50^{th} percentile data represents the mean (SD).

	Percentiles: Hip rotation and Extension						
	5th	**10th**	**25th**	**50th**	**75th**	**90th**	**95th**
Total Rotation	44°	46°	53°	59(11)°	66°	75°	82°
Internal Rotation	12°	15°	20°	26(8)°	31°	37°	42°
External Rotation	19°	23°	28°	34(9)°	40°	46°	50°
Hip Extension	+8°	+6°	-2°	-1.5(6)	-5°	-8°	-14°

6 week treatment protocol, participants were recruited who demonstrated hip mobility of less than the 50^{th} percentile, ideally in both directions. In total, approximately 250 men were measured in an attempt to find participants who fit the criteria. Given the difficulty of finding men who had limited mobility in both directions, and the fact that the study design allowed the participants to be compared to themselves (pre and post- treatment regime), there were some participants who demonstrated either marked limitation in one direction, but near normal in the other (e.g. subject # 4), or unusual asymmetry side to side (subject # 6) who were allowed to participate. See Appendix 6-1 for anthropometric data and ROM measurements for each of the participants. In total, 27 participants took part in the study. Two dropped out due to other commitments, and one due to illness, resulting in a total of 24. These were randomly assigned to 4 groups for treatment protocol. The exception to this was that those participants who worked off-campus, or admitted their reluctance to be compliant with an exercise regime, were placed into a control group, which only necessitated 2 sessions, as

opposed to weekly visits and home exercises. Of the 24 participants, 14 were undergraduate students (Kinesiology: 5, Engineering: 3, Other: 6), 9 were graduate students (Kinesiology: 8, Engineering: 1), and one participant worked off-campus. Table 6-2 outlines the primary physical activity that each of the participants regularly took part in, although many were active in more than one activity on a less focused level.

Table 6-2: Primary physical activity for each of the participants in the 4 treatment groups.

Group #	soccer	powerlifting	track & field	ice hockey	American football	general fitness	none
1	1	2	1	1	1		
2		1	1	3		1	
3			3	1	1		1
4			2	1	2	1	

6.2.2. Hip ROM measurements

Hip extension measurements were collected in supine lying, using the modified Thomas test (MTT) with the investigator controlling for ab/adduction and rotation as follows: the participant lay on his back. With the investigator's hand under the participant's lumbar spine, the investigator would flex the participant's hips/knees until she and the participant both agreed that the lordosis had reduced to a neutral position, indicating posterior rotation of the pelvis in the sagittal plane. The investigator's hand was then removed from the low back, and a blood pressure cuff replaced it, (while returning the spine to the same approximate position) with the cuff then being inflated to 60 mmHg. This pressure was monitored as one of the participant's legs was lowered leg passively by the investigator to a position of maximum hip extension without associated changes in pelvic position/pressure, as monitored by the BP cuff. The opposite leg was held passively in a position of hip/knee flexion by the investigator, which maintained the BP cuff at approximately 60 mm Hg. Participants were encouraged to give feedback as to their perception of pelvis position, in an attempt to further minimize pelvic rotation during hip extension.

Hip rotation measurements were captured with participants in prone lying. Bilateral internal rotation measurements were done simultaneously, as participants were asked to let both lower legs fall out to the side, while maintaining the knees at 90° of flexion. External rotation required the leg of interest to passively rotate across the midline. Pressure was applied on the ipsilateral pelvis by the investigator to ensure pelvis rotation did not occur. In those cases where large amounts of hip ER was present, the non-tested leg was abducted approximately 10° to allow free motion of the tested leg (Barbee Ellison et al., 1990).

Extension and rotation measurements were collected using a standard goniometer modified with the addition of two spirit levels: one on each of the arms, to improve accuracy of determining horizontal and vertical positioning. Every participant was positioned passively into hip extension and rotation by the principal investigator, who has over 30 years' experience as a clinical physiotherapist. Measurements were obtained by an assistant without sharing the results with the investigator at the time. Single measurements were taken, as the literature shows that single measurements can be as reliable as the average of multiple measurements (Boone et al., 1978; Bovens et al., 1990; Lea & Gerhardt J.J., 1995). Order of measurements was randomized.

6.2.3. Collection Procedure

After being initially screened for hip mobility appropriate for the study, participants were scheduled to return for their initial intake session. Anthropometric measurements were taken: height, mass, arm length, leg length, and the hip ROM measurements were repeated. Participants were encouraged to practice using the elliptical trainer to ensure comfort and coordination with its use. Reflective infra-red markers were attached to the body as described below, following which a calibration pose was collected with the motion capture system.

The first activity the participants were asked to do was to actively extend their hip to their perceived maximum while in an upright standing posture. They were given minimal guidance as to how to perform the action, other than to attempt to keep their upper body erect (i.e. avoid leaning the trunk forwards). Each activity was performed twice on each leg. Secondly, they were asked to complete a forward lunge: from their standing position, the floor was marked at a distance 1.5X their shin length. They were asked to step forward with one foot, until their toe reached the floor marking, and lunge down into forward hip flexion (backward hip extension) as low as was comfortable, while keeping their upper body erect. Again, this was repeated twice on each leg. The third functional trial was a “twist and reach” activity: two poles were set up aside the participants (in the frontal plane). The distance between the poles was 110% their body height, with the participant standing in the middle, feet shoulder width apart. Small knobs on the poles were secured at approximately the height of the person’s waist. They were asked to reach around and touch the knob on the right pole with their left hand, and the left pole with the right, without moving their feet.

Participants were subsequently asked to use the elliptical trainer. They chose a self-selected speed, with the instructions being, “choose a speed you would feel comfortable using for 30 minutes if exercising in the gym”. Although the stride length, hand position and speed were varied at the time, the results being discussed in this study utilized the 26” stride length, a speed 30% faster than self-selected, and their hands holding onto the oscillating handles of the elliptical trainer. Once they were up to speed and appeared comfortable with the activity, the motion was twice captured over approximately 4 cycles each.

6.2.4. *Motion Capture*

Vicon MX Motion System and Nexus software (Vicon Motion Systems, Oxford, UK) were utilized for capturing motion. Eight infra-red cameras collected data at a frequency of 60 Hz. Rigid plates with 4 reflective markers on each were attached via elastic straps to body segments bilaterally as follows: shin, thigh, foot, hand, forearm, upper arm, and overlying the midline on the posterior pelvis, forehead and over the T12 spinous process. In addition, single markers for calibration purposes only were attached over the posterior Rt scapula, C7 spinous process, sternal notch, and bilaterally over the medial and lateral aspects of each ankle, knee, wrist, elbow, ASISs, PSISs, greater trochanters, acromions, and earlobes.

6.2.5. *Treatment Protocol*

Participants were divided amongst 4 groups:

1. Hip flexibility
2. Hip flexibility and spine stabilization (motor control)
3. Spine stabilization and strengthening
4. Control

Groups 1 – 3 were scheduled for treatment sessions once a week for 6 weeks, starting after the initial data collection session. At that initial treatment session, exercises were determined based on which group they had been assigned to, and their current hip flexibility, trunk endurance and ability to disassociate their hip and spine movements. Thus, the exercises were individually tailored for each participant. They were asked to do their exercises at home at least 4 days per week, and were provided with a log book to assist with keeping track of the exercises. At no time were the exercises to reach a level where they required more than 30 minutes of their time per day. All participants were instructed not to change their normal exercise routine, other than to add in the new protocol. That is, if they already exercised daily, this routine was to be added to it.

Group 1: The philosophy behind the stretching programs included:

- Exercises would be more likely to be adhered to if they could be completed alone, i.e. no additional person was required as an assistant.
- A more thorough stretch might be obtained if stretching included not only the hip joint, but the entire side of the body which was undre stretch at the hip. For example, stretching of the right hip flexor would also include elevation of the right arm overhead, with extension and left side bending of the torso.

Figure 6-1: Whole body stretch of the right hip to increase ER

- Stretches were to be done in a position deemed to be one of hip function for each participant. Thus, stretching to increase right hip external rotation was accomplished in upright standing, with the hip externally rotated and knee flexed approximately 90°. The foot was anchored onto a table or other immobile object to secure its position. The participant was then instructed to twist the pelvis and upper body to the left, elevate the right arm, and attempt to apply overpressure through the trunk to maximize hip ER (Figure 6-1).

- A combination of static stretches (30 second hold), and ballistic (bouncing) stretches might be more beneficial than either one alone. Consequently, keeping these philosophies in mind, participants were given a stretching program that was unique to their tightness patterns. Hip extension stretches were generally done in the same position as the MTT, as well as a lunge position. Hip IR and ER were stretched in upright standing, as mentioned above, or in supine lying with the leg of interest rotated to the side (see Figure 6-2). Participants were encouraged to alter their positions if maximal tension across the hip joint could be increased by moving their arm, twisting their torso, etc.


Figure 6-2: Hip ER stretch in supine.

In addition to the home exercises, participants were expected to attend one stretching session per week, where their exercises were reviewed, further customized, followed by approximately 20 minutes of passive stretching by the investigator. Again, both static and ballistic stretches were used, with the concept of fascial chain connections being applied (Myers, 2001), thus arm and torso positioning being used to maximize tension in the side of the body being stretched. For example, as shown in Figure 6-3, while overpressure was being applied


Figure 6-3: Manual stretching to increase hip ER. Note that the hip is also put into extension, the torso twisted to the left, and the right arm is elevated, to maximize tension in the entire right anterior part of the body.

to stretch the right hip internal rotators (and thus increase ER), the participant was also positioned with hip extension, torso twist to the left and right arm elevation, all in an attempt to maximize right anterio-lateral body soft tissue tension.

Group 2: This group also received stretching similar to those in Group 1, but in combination with exercises aimed at improving hip/spine disassociation: maximizing active hip motion while minimizing concurrent spine motion. Generally, most participants began with supine bent knee fall-outs (BKFO), prone lying hip rotations, and upright standing unilateral hip motions (flexion/extension, circumduction, ab/adduction), with instructions to perform the hip motion actively but minimize associated spine motion. Depending on their ability level, these exercises were upgraded to increase the moment that the leg was applying to the trunk (e.g. supported BKFO progressed to unsupported, then repeated with a straight leg, and perhaps with a weight added to the ankle). For more complete descriptions of the exercises, see Appendix 6-4.

Group 3: Core strengthening and endurance combined with hip/spine disassociation was the goal for this group. In addition to the BKFOs and active hip motions, there was a strong focus on maximal activation of the trunk musculature. They received no instruction or exercises aimed at stretching the hip joint. In their initial treatment session, endurance tests were performed using the following positions:

- Plank: the torso/legs remaining in a straight line, with weight being borne on the forearms and toes.
- Side bridge: torso/legs in a straight line, weight being borne on lower forearm and the side of the feet (uppermost foot in front). This was repeated for both right and left sides.

- Biering-Sorensen position: lower half of body supported on a treatment table, securely held down with a non-elastic strap as well as manual pressure on the calves by an assistant. The upper body was unsupported, being cantilevered over the end of the table. With the arms crossed in front the chest, the participant was to hold the upper body in a line horizontal to the lower body.

For all tests, a meter stick with a sliding wooden caliper was used to monitor initial position of the elevated body part, to objectively determine when the participant was no longer able to hold the position.

In addition to the motor control exercises prescribed to group 2, this group also received instruction in the classic “bird-dog” exercise, planks, side-bridges, and supine lower abdominal exercises (see Appendix 6-4 for more detailed description of exercises). Exercises were progressed once the participant was able to complete a set number of repetitions (usually 20), without losing control of their neutral spine posture. Progressions involved adding weights to the moving limbs (arm or leg), increasing the scope of arm or leg movement (in the bird-dog or side-bridge), and in some cases, adding in more intense strengthening exercises such as the “suitcase walk”, or using the Body-blade™ in a vertical two-handed, side to side oscillation pattern (Moreside et al., 2006). Those participants who regularly used a gymnasium for fitness were advised how to tailor their workout at the gym to improve core strengthening.

Group 4: This was the control group. They attended an initial data collection involving the elliptical trainer and movement trials, then returned again 6 weeks later for a repeat data collection. They were instructed to not change their normal exercise routine during this time.

6.2.6. Re-test

The final re-assessment was scheduled after the participant completed approximately 6 weeks of their exercise routine. If they had been unable to exercise for one of the weeks for any reason, the final re-test was postponed one week, to allow a full 6 weeks of exercise. This appointment was basically a repeat of their initial visit: hip ROM was measured, as per the first visit. Again, the participants were all positioned by the single investigator and ROM measured by an assistant who was not aware of which treatment group they had been assigned to. This was followed by motion capture of active hip extension, lunges and twist/reach, in addition to the elliptical. The only cue that the participants were given at this time was, "You have just spent 6 weeks focusing on a specific exercise routine. Try to use it." They were given no hints regarding trunk bracing or movement, other than to keep the trunk rather erect, as had been the instruction at the intake.

For those who had been in group 3, endurance re-testing was scheduled as soon after this final elliptical/movement re-test as was possible. It did not take place on the same day, as it was felt that the fatiguing trials would affect the elliptical/movement performance, and vice versa. One participant was unable to attend for his final endurance testing as he was working out of town. He, therefore, sent his endurance results to us, but was unable to test his back extension endurance as he had no assistant to hold his lower body. All participants were not told their intake endurance values until after completion of the final re-test.

6.2.7. Kinematics

Marker data was initially processed using the Vicon Nexus software, then exported to Visual 3D (C-Motion Inc., Kingston, Canada) for further processing. 3-dimensional lumbar and hip angles relative to the pelvis were calculated using a Visual 3D algorithm with a

Cardan sequence of rotations. Joint angles were filtered with a 6 Hz dual pass Butterworth filter. Signals were screened for abnormalities, processing errors, and marker movement. For the elliptical trials, maximum and minimum joint angles were taken from the entire capture time, unless the signal drifted over time due to body position changes (i.e. neck flexion, which tended to increase lumbar flexion), in which case the max/min were extracted from a complete cycle deemed representative of the normal scope of motion. To calculate average joint positions for lumbar and hip flexion angles, trials were clipped to ensure complete cycles of motion (similar to heel strike to heel strike in gait studies). Symmetry was assumed in the elliptical and twist trials, and the right leg was used for statistical analysis. For the lunge and hip extension trials, joint angles were calculated at the instant where relevant peak joint motion occurred: that is, for a left leg forward lunge, spine and right hip angles were calculated at the moment of peak left hip flexion. Similarly during the right hip extension trials, angles were calculated at the instant of peak right hip extension. For the twist conditions, maximum and minimum angles were calculated based on the entire trial.

6.2.8. Statistical Analysis

Initial analyses utilized the SPSS (version 17) package with a significance level chosen at $p < 0.05$. Subsequent post-hoc tests on within-subject factors were conducted using SAS (version 9.2).

Hip Stretching outcomes: Paired t-tests were used to compare right and left sides for each of the dependant variables (hip extension, IR, ER, TotR) in the pre and post-treatment conditions. If no significant differences were found, these data were collapsed. Repeated measures ANOVAs were conducted on hip ROM for each of the measurements, using a within subject factor of pre/post treatment and between subject factor of treatment group.

Tukey's post-hoc tests were applied to any significant results.

Endurance outcomes: Paired t-tests were used to compare pre and post-treatment endurance times for the plank, right side bridge, left side bridge, and Biering-Sorensen test. Bonferroni adjustments were applied. Cohen's d effect sizes were calculated.

Spine and Hip angle changes pre and post-treatment: A series of repeated measures analysis of variance were performed for each dependant variable (various spine and hip angles), using a within subject factor of pre/post treatment and between subject factor of treatment group. One way analysis of variance was also performed on the difference between pre and post-treatment angles. Paired t-tests were conducted on individual pairs of pre and post results of interest, with Bonferroni adjustments.

6.3. Results

The results are initially presented for the 4 treatment groups based on hip mobility, core endurance, and motor control exercise progression. Following this, comparisons are made as to how these programs affected their movement patterns.

6.3.1. Hip ROM, Groups 1 and 2

When collapsed across all treatment groups, there were no significant differences between right and left hip extension, IR or ER (p = 0.375, 0.252, 0.060 respectively) However, the TotR did demonstrate a significant difference between sides (p = 0.005), thus were further analyzed separately.

There was a significant difference between the measurements taken before and after the 6 week treatment protocol for all measurement factors when collapsed across the treatment groups. Significance values were as follows:

- extension: p = 0.005

- IR, ER, Right TotR and Left TotR: $p < 0.001$

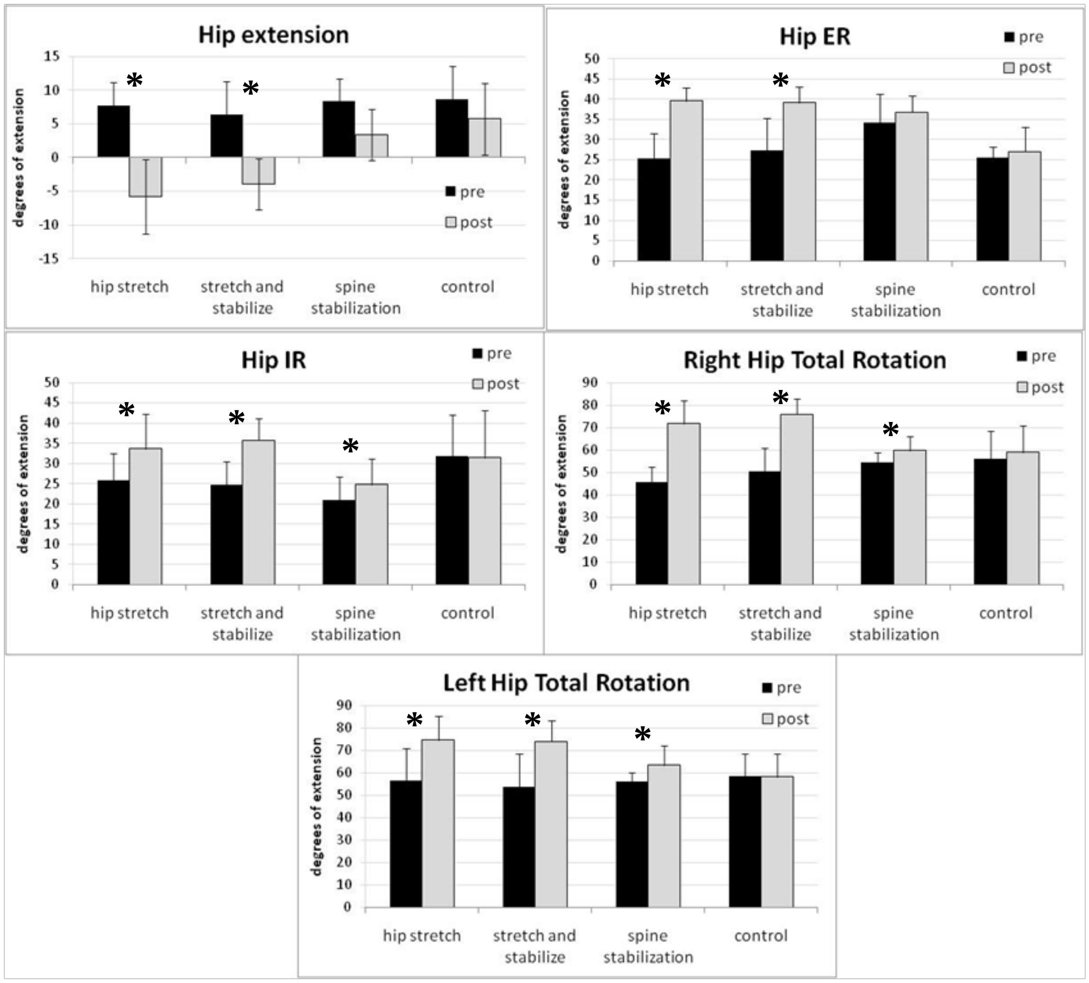


Figure 6-4: Average ROM in each of the treatment groups, for each direction of motion analyzed, comparing measurements before and after a 6 week treatment protocol. n = 6 participants per group (24 total). * = significant difference between the pre and post measurements ($p < 0.05$)

Post-hoc tests revealed that the two groups that received hip stretching increased their ROM significantly in all measurements, while the group that received only spine stabilization exercises also increased ROM significantly in hip IR and both rotations. The control group demonstrated no significant changes in ROM after 6 weeks (Figure 6-4, Table 6-3). The large increases in ROM in groups 1 and 2 are akin to changing from the 10th to the 75th percentile for extension, and the 20th to the 90th for total rotation. Equally important, every participant increased his ROM in each direction (see Appendix 6-2 for individual ROM results).

Table 6-3: Hip ROM measurements in degrees for all treatment groups before and after the 6 week stretching protocol. Extension is measured relative to the horizontal, thus a negative number indicates greater ROM. n = 6 participants in each group. *indicates significance at the 0.05 value

	treatment group	pre	post	difference	*p*- value
Ext	1	7.8(3)	-5.8(6)	-14	< 0.001*
	2	6.4(5)	-3.9(4)	-10.3	< 0.001*
	3	8.3(3)	3.4(4)	-4.9	0.067
	4	8.7(5)	5.8(5)	-2.9	0.264
ER	1	25.3(6)	39.5(3)	14.2	< 0.001*
	2	26.8(8)	39.2(4)	12.4	< 0.001*
	3	34.2(7)	36.8(4)	2.6	0.261
	4	25.6(3)	27.0(6)	1.4	0.527
IR	1	25.9(7)	33.6(9)	7.7	< 0.001*
	2	24.7(6)	35.8(5)	11.1	< 0.001*
	3	21.0(6)	24.7(6)	3.7	0.022*
	4	31.8(11)	31.5(12)	0.3	0.847
R TotR	1	45.8(7)	71.7(10)	25.9	< 0.001*
	2	50.3(10)	76.0(7)	25.7	< 0.001*
	3	54.4(4)	59.7(7)	5.3	0.048*
	4	56.2(12)	58.9(12)	2.7	0.283
L TotR	1	56.5(14)	74.6(10)	18.1	< 0.001*
	2	53.5(15)	73.8(9)	20.3	< 0.001*
	3	55.9(4)	63.3(9)	7.4	0.019*
	4	58.5(10)	58.1(10)	-0.4	0.888

6.3.2. Core endurance (Group 3)

This represents participants in group 3 only, who were tested before and after a 6 week strengthening/ endurance protocol, but received no flexibility stretches. As shown in Figure 6-5 and Table 6-4, trunk muscle endurance improved for all 4 test positions, but given the stringent significance level of 0.0125 calculated

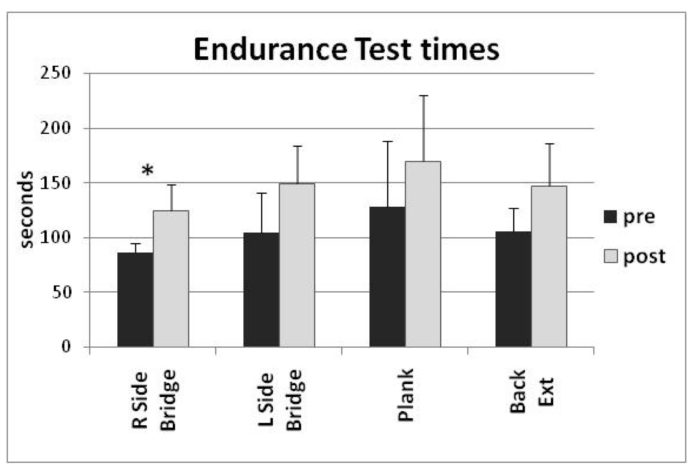


Figure 6-5: Endurance measured in seconds for the 4 test positions as labeled. * = significance at the 0.0125 level. (n = 6)

with the Bonferroni adjustment, only significantly so for the right side bridge (p = 0.006, 0.039, 0.018 and 0.039 for RSB, LSB, plank and back extension respectively). Likely the lack of significance is due to the low number of participants (6) in this group, resulting in statistical power levels ranging from 0.29 to 0.47 for the three non-significant factors. The effect sizes were 2.09, 1.26, 0.6 and 1.36 (same order as above), indicating that despite lack of statistical significance, the effect of the exercise program on endurance was large for all but the plank exercise, which could still be considered moderate. On average, the endurance times improved between 38% and 53% for all 4 tests. See Appendix 6-3 for individual endurance times for each position.

Table 6-4: Average endurance outcomes from group 3 (n =6), comparing their endurance times before and after a 6 week strengthening/endurance protocol. The bottom row represents published norms collected over a group of 92 males, mean age 21 yrs.

	R side bridge	L side bridge	Plank	Back ext
pre	87 (8)	104 (37)	128 (60)	105 (22)
post	125 (24)	150 (35)	170 (70)	147 (39)
Published norms - McGill (2002)	95 (32)	99 (37)	_______	161 (61)

6.3.3. Exercise Progression (Groups 2 and 3)

In addition to the stretches (as discussed above), those participants in group 2 received motor control exercises focusing on active hip motion with minimal associated spine motion. Their exercise program began at a level where they could demonstrate control of the lumbar spine during active hip motions, and was progressed according to their level of spine control ability. Every participant was able to progress to a greater level of difficulty throughout the 6 week protocol. The focus was on motor control, as opposed to endurance. Table 6-5 shows the exercises that each of the participants in groups 2 and 3 were able to progress through.

Table 6-5: Exercise progression for groups 2 and 3. * = beginning point of exercise regime, and ** = an exercise progressed from one that was already in their routine. Exercises progress in difficulty as the columns move left to right within the darker lines. Participants in group 2 were concurrently being given stretches to do at home, while those in group 3 only focused on core/trunk stability, thus were generally able to progress to a higher level of difficulty.

subj #	Rx grp	classic bent-knee fallouts (BKFO)	BKFO with legs unsupported	unsupported BKFO with straight leg	weights added to ankle	prone lying hip rotation	lower abs with unilateral leg lowering	lower abs with sequential leg lowering	lower abs with bilateral leg lowering	lifting head and shoulders added	single leg stance: hip patterns	lunges, etc with stable pelvis	elastic resist hip motion in single leg stance	side bridge	side bridge with weight in upper hand	side-bridge with body rolling to opposite SB	bird dog: diagonal lifts	bird dog: arm/leg "squares"	weights added to ankle or hand	weights placed on back	plank	plank into side bridge	one armed plank	suitcase walk	body-blade
1	2	*	**	**		*	*	**			*	**	**												
6	2	*	**	**		*		*			*	**	**												
9	2	*	**	**				*			*	**	**	*			*								
14	2		*	**							*	**													
32	2	*	**	**							*	**			*							*	**		
4	2		*	**				*			*	**													
29	3	*	**	**	**		*	**	**	**	*	**		*	**		*	**	**	**	*		**	**	
31	3	*	**	**			*							*	**		*	**			*	**			
35	3		*	**	**			*	**		*	**			*		*	**							
34	3		*	**	**	*					*	**			*	**		*	**	**				**	**
37	3		*	**	**	*	*	**	**		*	**	**		*	**		*	**				**	**	
43	3	*	**	**		*			**	**	*	**		*	**			*	**			**			

6.3.4. <u>Functional hip mobility</u>

Despite the large increase in passive hip mobility in groups 1 and 2, there were no significant increases in the amount of hip extension or rotation utilized during the functional movement tests. In fact, the average amount of hip extension in these two groups actually decreased in the active hip extension trials (Figure 6-6), while changing less than 1° in the elliptical and lunge trials. Similarly, the amount of hip rotation utilized in the elliptical and twist/reach trials also decreased in the post-stretching trials (Figure 6-7).

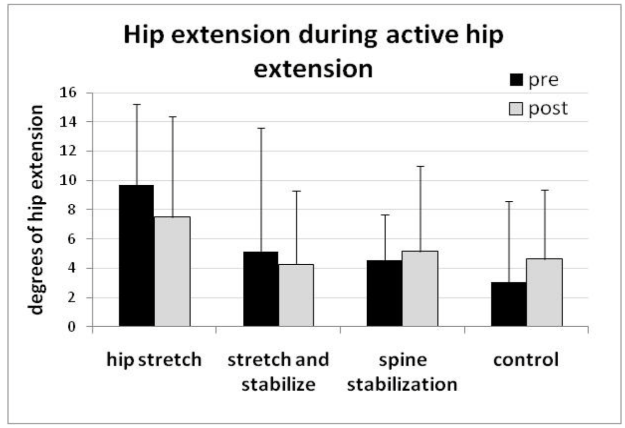


Figure 6-6: Degrees of average right hip extension utilized in an upright standing trial, when asked to actively extend the right hip. (n = 6 per treatment group)

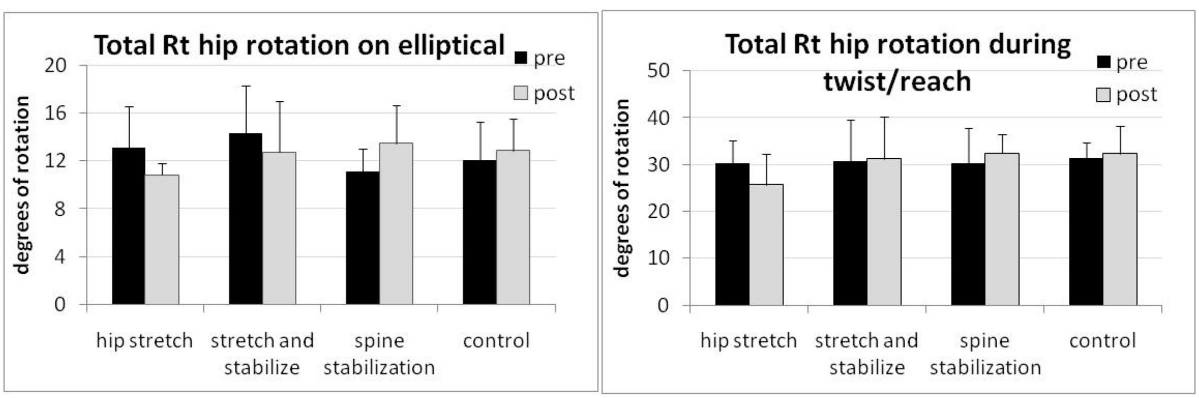


Figure 6-7: Degrees of average total hip rotation (IR + ER) utilized during use of the elliptical trainer, as well as a movement trial involving twisting and reaching sequentially from one side to the other in standing. (n = 6 per treatment group)

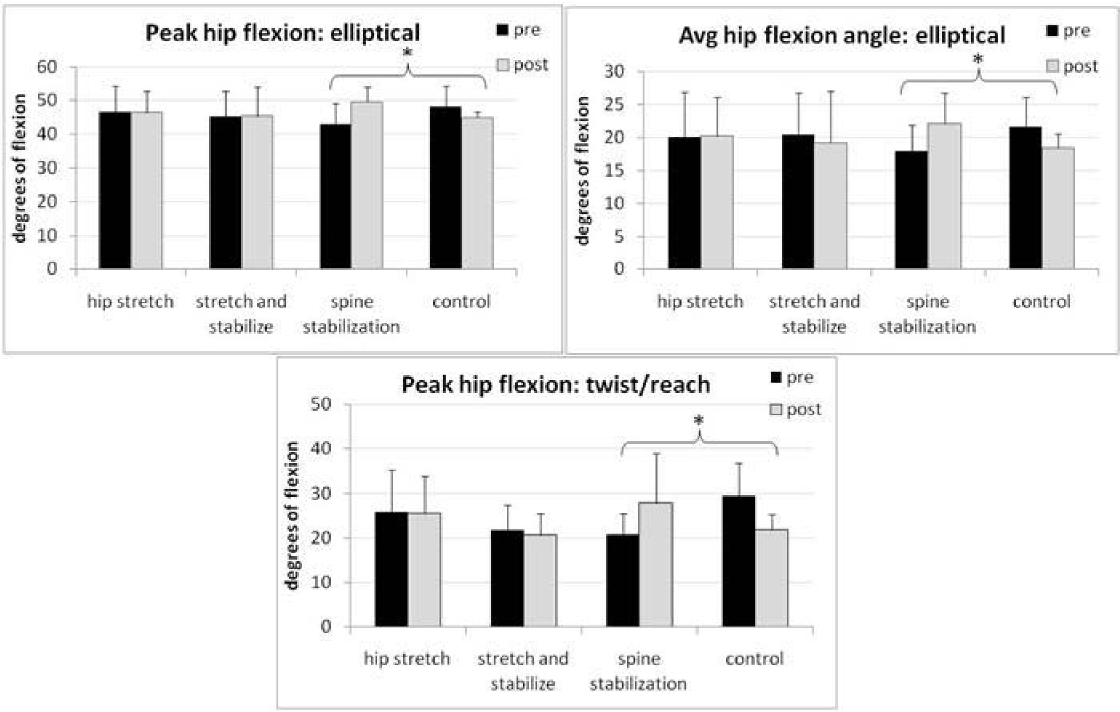


Figure 6-8: Examples where the spine stabilization group demonstrated a significant increase in hip flexion angle relative to the control group. * = significant at the 0.0125 level. (n = 6 per treatment group)

It is worth noting that group 3, who received spine stabilizing exercises only, did demonstrate a significant increase in peak hip flexion and average hip flexion angle (mean over complete cycles) on the elliptical trainer, as well as the twist/reach condition when compared to the control group (Figure 6-8). Using one-way ANOVAs to look at the *differences* between pre and post-treatment hip flexion across the different treatment groups, *p*-values of 0.012, 0.036 and 0.007 were calculated for peak and average hip flexion on the elliptical and peak hip flexion in twist/reach, respectively. Tukey's HSD post-hoc tests indicated significant differences between groups 3 and 4 in each condition (p = 0.008, 0.026 and 0.004, respectively). In these trials, the stabilization group increased their hip flexion by 6.5°, 4.1° and 7.2°, whereas the control group decreased hip flexion by 3.2°, 3.2°, and 7.7°, for peak and average hip flexion on the elliptical and peak hip flexion in the twist/reach, respectively.

6.3.5. Functional spine motion

For groups 2 and 3, the exercise focus had included core stabilization and spine/pelvis motor control. There was only one instance where lumbar motion was significantly different post-treatment across all groups: concurrent lumbar rotation was significantly less post-treatment during the active hip extension movement (p = 0.015) (Figure 6-9). Of specific interest is that the group who received both stretching and stabilizing demonstrated the greatest decrease in spine motion in this condition, changing from 7.4(3.8)° to 3.8(3.6)°. Otherwise, there were no other significant differences for spine motion when comparing spine flexion/extension, rotation or side bending post-treatment to that demonstrated pre-treatment.

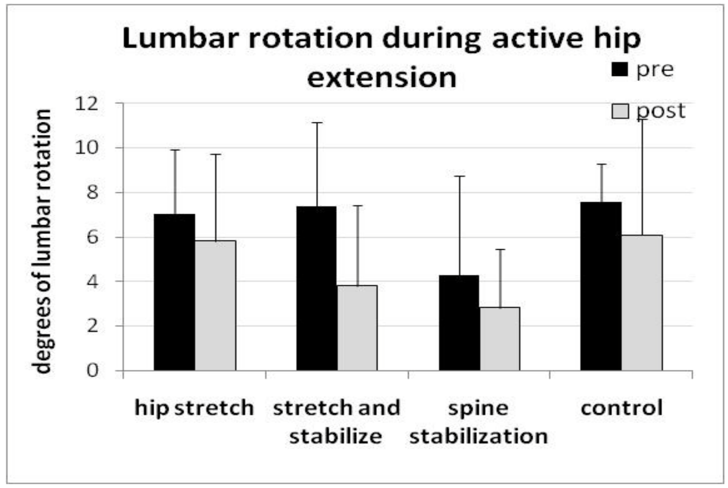


Figure 6-9: Concurrent lumbar rotation associated with active right hip extension. Lumbar motion was significantly less post-treatment (p = 0.015). (n = 6 per treatment group)

The hip stretching group (group 1) demonstrated the largest *increase* in the amount of lumbar extension used in the lunge and active hip extension trials, significantly so in the active hip extension trial (p = 0.001), where their lumbar extension increased from an average of 9.2 (4.8)° initially to 15.1(3.9)° post-stretching (Figure 6-9). Thus it appears that they did not differentiate between hip extension and lumbar extension when attempting to extend their leg behind the body.

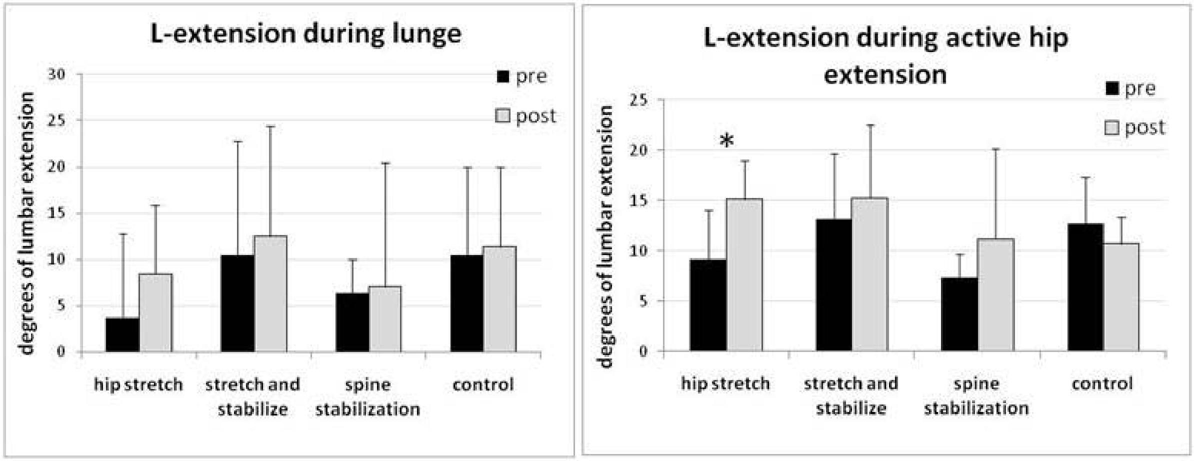


Figure 6-10: Lumbar extension angles during a left leg forward lunge and active right hip extension trials. * = significant difference between pre and post at the .0125 level, using paired t-tests. (n = 6 per treatment group)

Despite the lack of significance, it is worth observing the graphs depicting lumbar motion in Figure 6-11 and Figure 6-12. Group 3, the spine stabilization group, repeatedly showed a decrease in lumbar rotation after the 6 weeks of exercises, both in the lunge and active hip extension conditions (Figure 6-9), as well as on the elliptical trainer. This is especially so when compared to the control and hip stretching groups. Similarly, on the elliptical trainer, the group that received both spine stabilization and stretching demonstrated a decrease in spine motion in all directions after the 6 weeks of exercises, whereas the stabilization only group improved in all but the side bending condition (Figure 6-12).

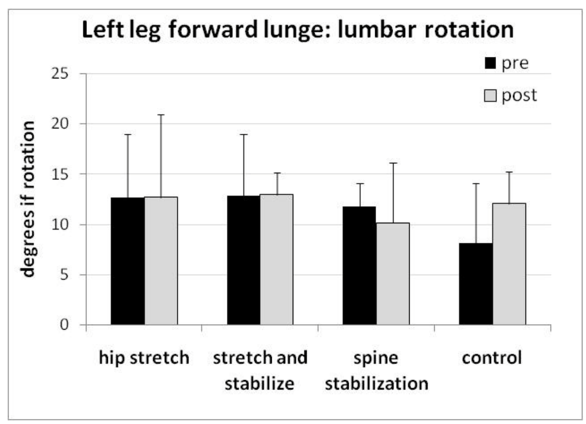


Figure 6-11: Total amount of lumbar twist pre and post-treatment, for the 4 exercise groups. (n = 6 per treatment group)

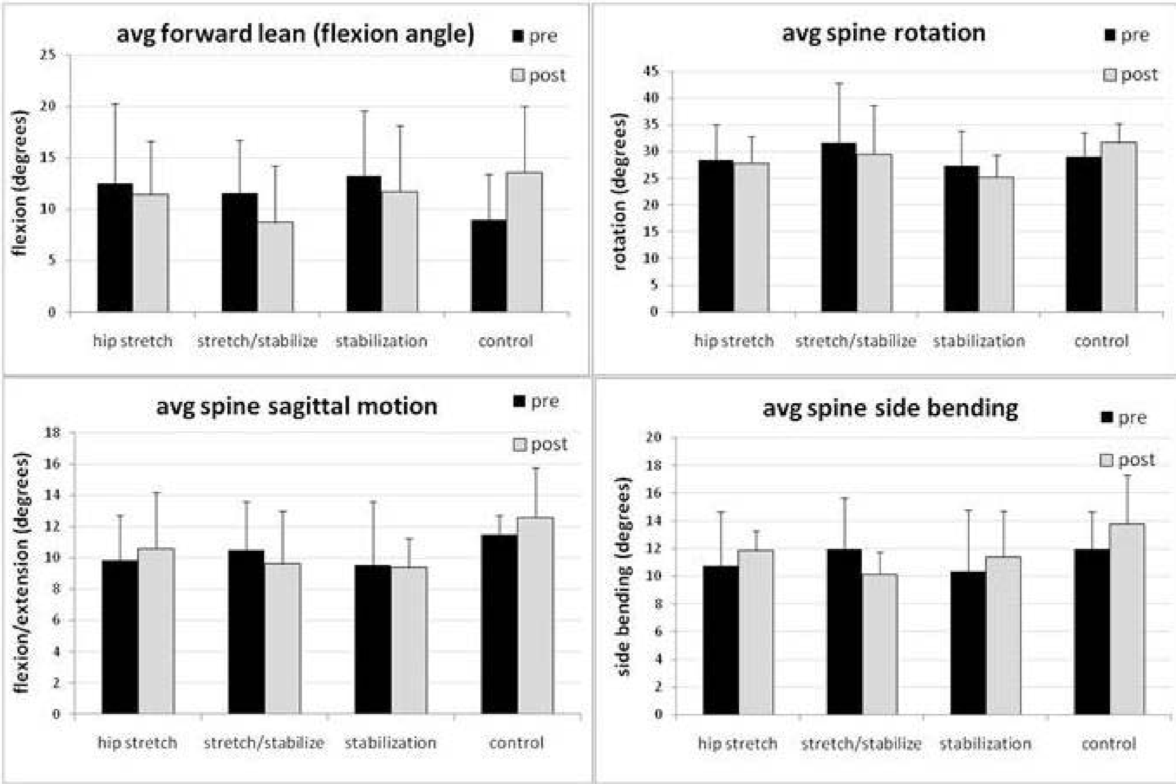


Figure 6-12: Spine motion while exercising on the elliptical trainer, averaged across 6 participants in each group.

6.4. Discussion

The combination of ballistic and static stretches used in this study resulted in large increases in passive hip mobility. Participants in the stretching groups changed from being in the lowest 20th percentile to the highest 75th in 6 weeks. This degree of improvement in hip extension is similar to the 12 - 14° documented by Winters et al (2004), also in a group of healthy young males with limited hip extension, using either passive or active stretches. They did not, however, include hip rotation measurements in their study. There is no scientific literature describing the hip rotation stretches used in this study. They were individually tailored to each participant, depending on his pattern of restriction, while maintaining the hip in a relatively neutral sagittal position. The concept that fascia adheres to muscles, transmits force and acts as a connecting structure between anatomically distinct muscles has been described (Huijing, 2007; Huijing, 2009; Maas, Jasper, Baan, & Huijing, 2003; Rijkelijkhuizen, Baan, de Haan, de Ruiter, & Huijing, 2005; A. Stecco et al., 2009; C. Stecco et al., 2006) thus the concept of myofascial stretch cannot be discounted. Further studies are warranted to determine if this type of stretch is superior to the typical style that would focus locally on a specific muscle or joint.

All participants in groups 2 and 3 improved their level of proficiency in core stabilization exercises. They were able to progress through levels which increased the challenge of moving the limbs independent of the trunk, by increasing the moment induced by the limb (i.e. leg straightening, added weights, elastic resistance to standing hip exercises). Only one participant did not improve in his back extension or plank endurance, but did improve 31% and 43% in his right and left side bridges, respectively (See Appendix 6-3). This particular participant was not a regular exerciser prior to the study, thus perhaps did not embrace the strengthening protocol with the same enthusiasm as those who were already striving for fitness. Many of those in group 3 were already physically active in weight-lifting and heavy resisted activity, compared to those in the other groups, which was purely a function of the random assignment. Thus, the addition of the "suitcase walk" and added ankle or wrist weights to further the challenge for these participants may be more than is expected of a less physically strong population.

A major finding in this study is that improved passive hip flexibility did not translate into increased range of hip motion utilized during functional activities. Despite the highly significant improvements in passive hip extension and rotation in groups 1 and 2, not a single instance of increased functional hip motion was demonstrated in the post-testing session. In fact, in many instances, participants tended to use less of the available hip joint range in the post-test, while demonstrating significantly more lumbar extension. This is evidenced when comparing Figure 6-6 to Figure 6-10: despite the fact that the hip had gained passive range of motion, attempts to utilize the new ROM in functional hip extension resulted in the exertion being misplaced to the lumbar spine. It would appear that they were not adept at distinguishing where the extension was occurring, just that they were focusing on getting the leg further behind their body. Consequently, attempts of active hip extension resulted in an increase in lumbar extension stress, as opposed to the spine sparing effect that was intended with the hip flexor stretching protocol. It appears that the concept of constraining lumbar rotation may be more readily incorporated into movement, perhaps because it provides greater visual feedback, and oscillates around an obvious mid-point of 0°. Constraining lumbar flexion/extension, however, was a much more difficult concept for the participants to incorporate into movement patterns, yet is one of the motions known to be injurious to the lumbar spine and discs (Callaghan & McGill, 2001; Drake & Callaghan, 2009; Tampier et al., 2007). This lack of correlation between ROM measured with the MTT and functional movement is in keeping with that found by Lee et al (1997), although their study was a cross-sectional design and did not involve measuring changes in mobility/motion. Caplan et al (2009) did document increased dynamic hip flexion after a 5 week stretching protocol aimed at hamstring flexibility in a group of male rugby players, but this was demonstrated during high speed running, where the hamstrings would be working strong eccentrically, thus a different role than our slower, less ballistic movement trials.

Much of the literature discussing changes in movement patterns subsequent to an exercise routine is sport specific (Grimshaw & Burden, 2000; Kiesel, Plisky, & Butler, 2009;

Lephart, Smoliga, Myers, Sell, & Tsai, 2007). In addition to basic stretches, participants in those studies practiced movements that were required of their sport, thus having more of a chance to "groove" new motor patterns, perhaps increasing the likelihood that the new motor patterns would transfer to function. In our study, the elliptical trainer and twist/reach motions were never practiced as a motor control exercise. Doing so may have improved participants' abilities to transfer new motor patterns to function. The lunge and active hip extension were both an exercise and a test, yet the only improvement detected in spine control in these two activities was seen in lumbar rotation, or twist. No significant improvement in control of flexion/extension was demonstrated after 6 weeks of intervention, despite the fact that these two motions were chosen to challenge sagittal control. The amount of hip flexion/extension measured varies with pelvis position, but unlike most other joint measurements, the sagittal position of the pelvis in upright standing is highly variable. In this study, a calibration pose at the beginning of each motion capture session was used to represent 0° of hip motion. Participants were asked to stand erect with their arms in approximately 90° of abduction. However, it was noted that some participants would increase their pelvic anterior tilt in this position, while others went into a posterior pelvic tilt. Therefore, the hip position of 0° would vary according to the preferred posture of the participant: an increase in anterior pelvic tilt would assign 0° to a relatively more flexed hip, whereas a posterior pelvic tilt would relatively extend the hip. Consequently, the variability in this measure, between subject and between day, may have been affected by these changes in pelvic orientation.

Limitations exist in this study. Healthy young adults were chosen for analysis in an attempt to reduce the likelihood that participants had reduced hip mobility due to arthritic change, or would have arthritic changes in the spine. The results may not transfer to an older or low back pain population. The effect of limited hip mobility in an equivalent female population should be investigated. Both sexes were not done at this time as it would have required twice the population, thus extended the time line immensely.

Having more participants in each group might have improved the statistical power, and thus significance. However, finding participants who demonstrated limited hip mobility in both directions tended to put them in approximately the lowest 10^{th} – 20^{th} percentile, necessitating screening of large numbers to find appropriate participants. The population at our university is often absent from the campus every second semester due to work terms. Consequently, there was a constant challenge to find participants, perform the intake and out-take assessments, and fit in a 6 week treatment protocol before exams and their leaving at the end of semester.

Only four tests were chosen as representative of functional movement: the lunge and active hip extension were chosen as a challenge to sagittal motion (and lumbar twist), while the twist/reach was to produce rotation in the hips and spine, as well as spine flexion. Other movements may have resulted in different outcomes.

As seen in the outcome graphs, variability in hip and spine movement is high. Despite attempts to standardize motion with movements relative to anthropometrics (lunge distance, twist/reach distance), it was noticed that people would move differently from trial to trial, let alone day to day. Similarly, inter-subject variability was high, as is demonstrated by the standard deviation bars. Yet to constrain movement further might have interfered with "normal movement". This variability likely hindered statistical significance, and would best be dealt with by increasing the number of participants (thus improving statistical power). There are many instances in this study where the control group demonstrated moderate differences in joint angles between sessions, when it would have been expected that their measurements should have remained fairly similar. This demonstrates the large variability in normal motion, day to day. Future research investigating changes in joint flexibility based on diurnal rhythms and daily activity would be of value.

Compliance to exercise protocols was based on feedback provided by the participants. Despite the fact that they filled out exercise logs, the onus was on them to provide truthful information. Most of the participants in this study were highly motivated to improve their strength and/or flexibility, perceiving it would improve their performance in some way. Weekly sessions with the investigator also tended to encourage them to adhere to the exercise protocol. In that some of the participants admitted to not exercising during some specific weeks (and thus their program was extended, as previously discussed), and that the improvements in ROM and/or endurance/motor control were considerable, it was felt that most participants were honest about their exercise compliance.

6.5. Summary

The main finding in this study is that large improvements to passive hip flexibility and spine stability made little difference to functional movements. The new mobility is not used, as participants deferred back to older, engrained movement patterns. This was especially so in the group that received stretching only, with no associated education re core stability: they demonstrated an increase in lumbar motion, and a decrease in hip range utilized during functional movements, thus indicating poor proprioceptive awareness as to where the extension pattern was occurring. Those participants who received spine stabilizing exercises (with our without associated hip stretching) tended to demonstrate better spine control, especially in the rotation axis (lumbar twist). Further research into this area of transference of mobility and/or strength to function is warranted. Stretching and strengthening programs form the base for many rehabilitation programs. It is important that we understand their applicability to functional movement, and how best to maximize this transference.

6.6. Appendix

<u>Appendix 6-1:</u> Participant anthropometrics and pre/post hip ROM

				PRE								POST							
subject #	**Rx grp**	**height**	**mass**	**RIR**	**LIR**	**RER**	**LER**	**R tot R**	**L tot R**	**Rext**	**Lext**	**RIR**	**LIR**	**RER**	**LER**	**R tot R**	**L tot R**	**Rext**	**Lext**
11	1	181	80	20	25	26	46	46	71	10	12	35	33	44	43	68	87	-4	-7
15	1	194	72	28	37	25	27	53	64	4	3	46	40	44	47	86	44	-15	-16
17	1	189	116	20	24	20	22	40	46	8	13	24	25	42	38	66	63	-6	-3
18	1	163	63	22	18	20	23	42	41	14	8	22	26	35	40	57	66	-7	-8
39	1	175	58	35	37	20	36	55	73	5	6	46.5	36.5	26.5	45	73	81.5	1.5	0.5
28	1	174.5	77	23	22	16	22	39	44	4	6	31	34	38	36	69	70	-3	-2.5
1	2	190	97	30	35	32	28	62	63	10	10	38	43	40	42	78	85	-4	-1
4	2	178	69	18	18	20	20	38	38	1	-1	32	31	33	31	65	62	-4	0
32	2	171	73	25	15	18	26	43	41	3	3	28	26	43	37	71	63	0	1.5
6	2	171	74	29	40	33	28	62	68	7	-2	37	40	44	36	81	76	-5	-5
9	2	171.5	102	22	25	20	20	42	45	14	12	39	39	38	39	77	78	-11	-10
14	2	176	80	27	22	30	32	57	54	9	9	40	36	44	43	84	79	-4	-4
29	3	179	79	13	13	43	40	56	53	6	4	14	19	41	39.5	55	58.5	3	2
34	3	184	95	21	23	26	30	47	53	6	9	21	25.5	32	29.5	53	55	0.5	6
31	3	184	80	30	30	30	40	60	70	7	8	30	40	28.5	38.5	58.5	78.5	12	9
35	3	172	52	25	20	26	35	51	55	9	2	20	19	36	39	56	58	0	-1
43	3	173	82	19	23	39	33	58	56	14	14	27.5	26	39	36	66.5	62	0	2
37	3	186	93	15	20	42	44	57	64	5	16	26.5	28	42.5	40	69	68	4	3.5
2	4	175	69	51	49	26	26	77	75	10	20	58	48	19	25	77	73	14	11
3	4	174	78	26	33	30	23	56	56	4	8	27	26	40	34	67	60	4	4
5	4	183	95	20	25	20	25	40	50	5	5	26	28	19	21	45	49	10	12
42	4	179	91	39	35	20	30	59	65	12	14	36	36	24	31	60	67	3	2
38	4	178	101	23	29	26	21	49	50	8	3	21	21.5	27.5	26.5	48.5	48	7.5	5
30	4	178	74	26	25	26	24	52	49	2	3	26	24	30	27.5	56	51.5	-2	-1.5

<u>Appendix 6- 2:</u> Hip extension and rotation ROM, pre and post-treatment

Only those participants who were in the stretching groups are listed in this table. Group 1 received stretching only, group 2 received stretching as well as core exercises focusing on motor control. Degrees of hip extension are relative to the horizontal.

subj.no	Rx group	Rext pre	Rext post	Lext pre	Lext post	RIR pre	RIR post	RER pre	RER post	RTotR pre	RTotR post	LIR pre	LIR post	LER pre	LER post	LtTotR pre	LtTotR post
11	1	10	-4	12	-7	20	35	26	44	46	79	25	33	46	43	71	76
15	1	4	-15	3	-16	28	46	25	40	53	86	37	44	27	47	64	91
17	1	8	-6	13	-3	20	24	20	42	40	66	24	25	22	38	46	63
18	1	14	-7	8	-8	22	22	20	35	42	57	18	26	23	40	41	66
39	1	5	1.5	6	0.5	35	46.5	20	26.5	55	73	37	36.5	36	45	73	81.5
28	1	4	-3	6	-2.5	23	31	16	38	39	69	22	34	22	36	44	70
1	2	10	-4	10	-1	20	38	40	40	60	78	35	43	40	42	75	85
4	2	1	-4	1	0	25	32	18	33	43	65	15	31	26	31	41	62
32	2	3	0	3	1.5	18	28	20	43	38	71	18	26	20	37	38	63
6	2	7	-5	-2	-5	29	37	33	44	62	81	40	40	28	36	68	76
9	2	14	-11	12	-10	22	39	20	38	42	77	25	39	20	39	45	78
14	2	9	-4	9	-4	27	40	30	44	57	84	22	36	32	43	54	79
average		**7.4**	**-5.1**	**6.8**	**-4.5**	**24.1**	**34.9**	**24.0**	**39.0**	**48.1**	**73.8**	**26.5**	**34.5**	**28.5**	**39.8**	**55.0**	**74.2**
st dev		4.2	4.4	4.8	5.1	4.9	7.8	7.1	5.3	8.7	8.6	8.5	6.5	8.4	4.5	14.2	9.4

<u>Appendix 6- 3:</u> Endurance time (seconds), pre and post-treatment

Those participants in treatment group 3 received an exercise program focusing on endurance as well as motor control of the trunk with regards to hip motion. Subject #43 was unable to report back extension endurance, as his final testing was done off-site with no assistant to stabilize his lower body.

subj.no	Rx group	RSB pre	RSB post	LSB pre	LSB post	Plank pre	Plank post	BackExt pre	BackExt post
29	3	80	100	85	125	85	140	75	127
34	3	90	105	92	150	75	150	100	120
31	3	80	105	70	100	100	95	115	115
35	3	90	140	170	170	200	260	135	205
37	3	80	140	83	200	100	120	100	169
43	3	100	158	125	152	210	253	---	---
average		87	125	104	150	128	170	105	147
st dev		8	24	37	35	60	70	22	39

Appendix 6-4: Descriptions of exercises

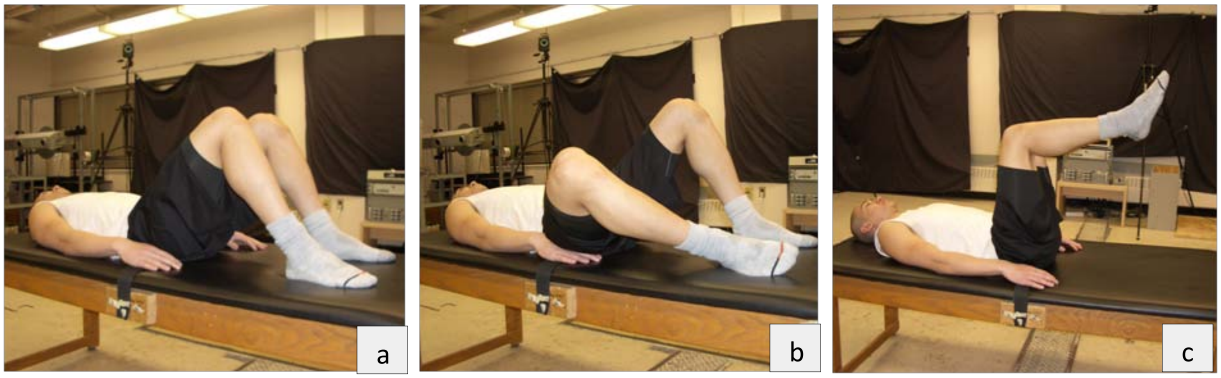


Figure 6-13: Bent Knee Fall-outs: a) Supine lying with feet supported, b) one leg at a time is allowed to rotate out to the side, while pelvic motion is kept to a minimum. c) progression involves legs being unsupported; weights may be added to ankles to further the challenge. (Sahrmann, S., 2002)

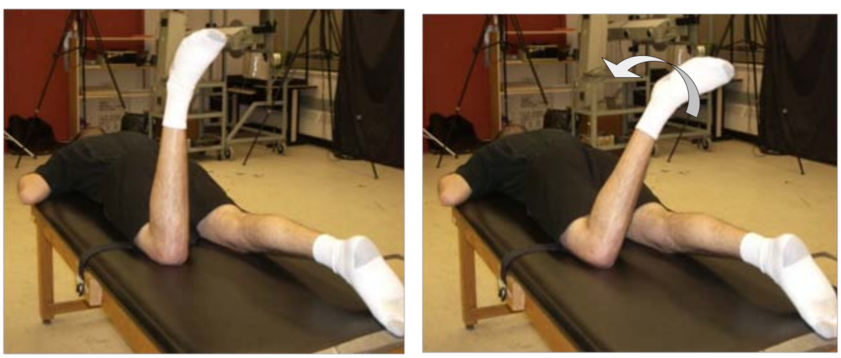


Figure 6-14: Prone lying hip rotation: Internal and external hip rotation, while concurrent pelvis rotation is kept to a minimum. (Sahrmann, S., 2002)

Figure 6-15: Unilateral and bilateral lower abdominal exercises: from a starting position of both hips being flexed (as in Figure 6-13c), one leg is slowly lowered to the plinthe, then raised back up. Lumbar spine must be maintained in a neutral lordotic position. Progression involves lowering both legs simultaneously. (Sahrmann, S., 2002)

Figure 6-17: Suitcase walk. Walking with 32 kg weight in one hand while maintaining an upright posture.

Figure 6-16: Hip abduction, extension, external rotation against elastic resistance, while minimizing associated spine motion.

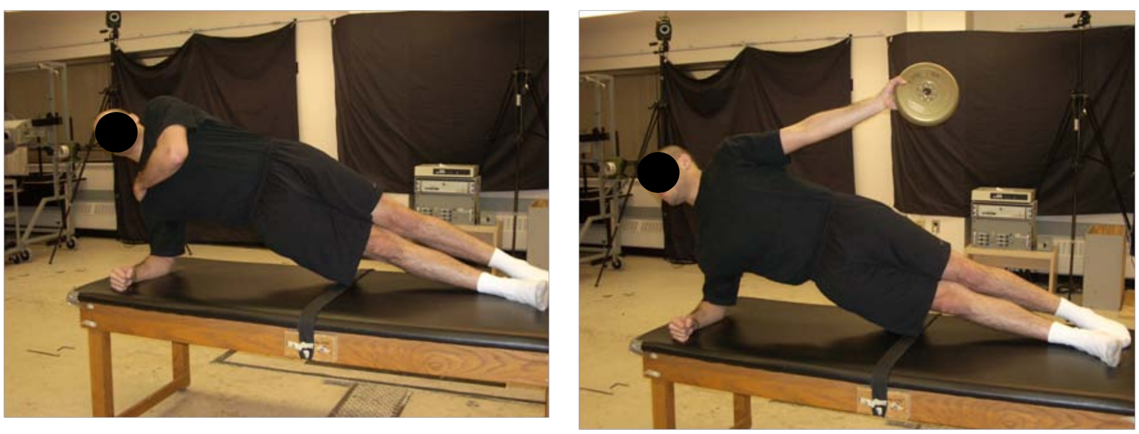


Figure 6-18: Side bridge, and side bridge with weight. Further progression would involve moving the weight in an anterior and posterior direction, to challenge rotational stability of the spine (McGill, 2002)

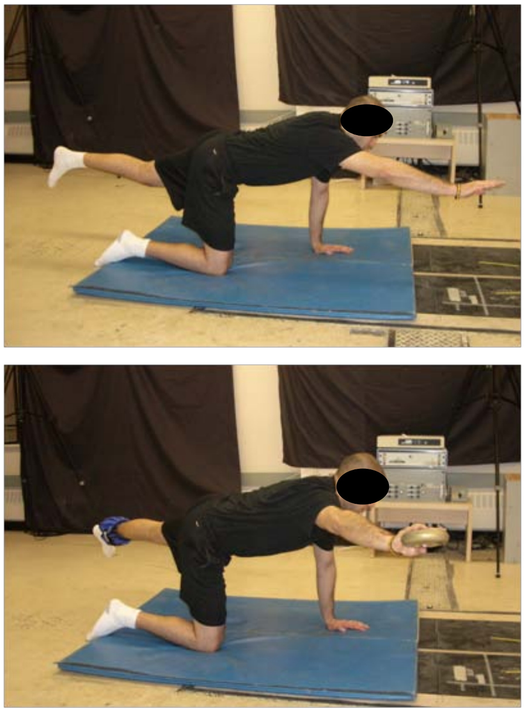


Figure 6-19: Bird-dog, progressing to abducting arm and leg, while maintaining spine in neutral position. Weights are added to the hand and ankle as an added challenge. (McGill, S., 2002)

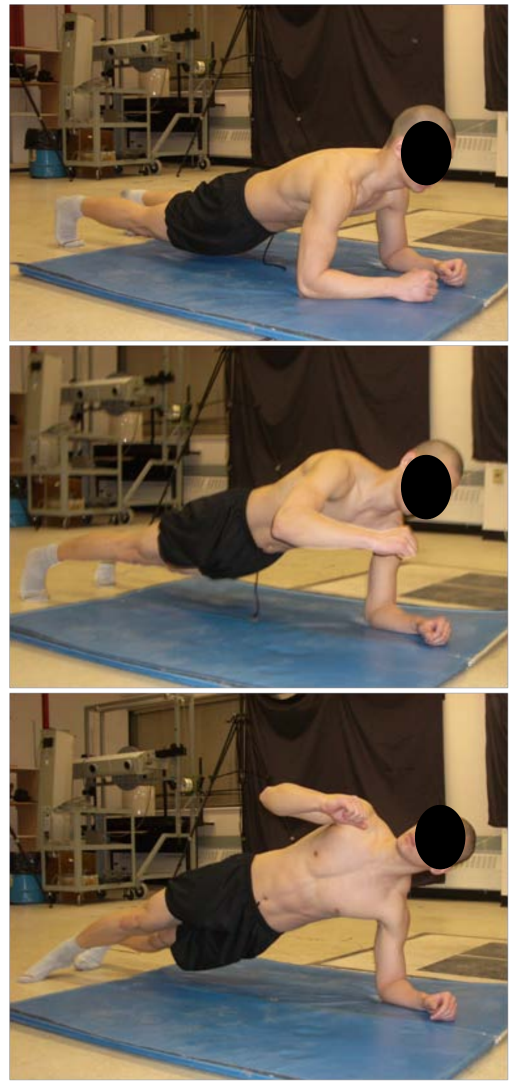


Figure 6-20: Plank with body roll; eventually rotating into a side bridge. Spine position controlled. (McGill, S., 2002)

7. Hip joint rotational stiffness: comparing groups in the 10th and 90th percentile

7.1. Introduction

Young adults with idiopathic limitation of hip rotation are often seen clinically, as are those with unusually large amounts of rotation. While most often the literature cites "average" range of motion (ROM), normative data, by definition, will also include those whose mobility is much more or less than the average. To date, little research has been dedicated to these subsets in a healthy, pain-free population. Evidence suggests that limited hip rotation often results in abnormal movement patterns in the lumbar spine, and associated low back pain (Barbee Ellison et al., 1990; Chesworth et al., 1994; Cibulka et al., 1998; Cibulka, 1999; Vad et al., 2003; Vad et al., 2004; Van Dillen et al., 2008). Thus, it is important to understand the mechanical properties of the soft tissues surrounding the hip joint, which are a major determinant of available ROM (Crawford et al., 2007; J. Hewitt et al., 2001; J. D. Hewitt et al., 2002).

Anatomically, the strong ilio-femoral, or Y-ligament, in front of the hip joint is believed to be the main structure limiting hip extension and external rotation (ER) (Gray, 1974; Hidaka et al., 2009). The maximal strain value of this ligament has been documented as 6.2% (J. Hewitt et al., 2001), considerably less than the 10.9% reported for the inferior gleno-humeral ligaments of the shoulder (Bigliani et al., 1992). Hip internal rotation is limited more so by the posterior ischio-capsular ligament (Gray, 1974). However, this posterior ligament has been shown to fail at far lower tensile stress than the anteriorly situated inferior ilio-femoral ligament (136(74.6) N compared to 351.3(159.4) N, respectively). Thus, the muscular contributions of piriformis, the obturator and gamelli groups as well as medial fibres of gluteus medius also play an important role in limiting internal rotation (IR).

The concept of a joint neutral zone (NZ) has been described by Panjabi (2003) as that part of the ROM within which there is minimal resistance to joint movement. Clinically, this is also referred to as the resting position of the joint, where the soft tissues constraining the joint are at their slackest (Gray, 1974). Perturbations applied to the joint when in this position would therefore be met with less passive resistance than if applied in other positions. Following Panjabi's analogy of a ball in a bowl, the question arises as to whether those people with highly flexible joints also demonstrate changes in the NZ? That is, as depicted in Figure 7-1, does the NZ stay approximately the same, but the stiffness of the constraining ligaments and soft tissues decrease (7–1b) or does the stiffness remain the same, but the size of the NZ increase (7-1c) , rendering them potentially less able to resist perturbations in this large slack region?

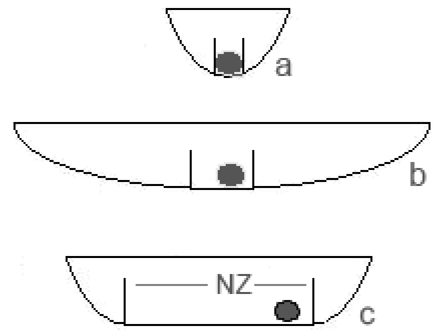


Figure 7-1: Depictions of 3 different scenarios of joint stiffness and neutral zone (NZ). a) joint with small neutral zone, stiff constraints b) joint with relatively small neutral zone, constraints not stiff c) large neutral zone, stiff constraints

The question arising then, is can the difference in mobility in these two groups be explained as purely a matter of soft tissue flexibility? That is, can it be explained solely by differences in the stiffness of the constraining structures, or does the width of the associated neutral zone differ between groups, leaving the flexible group with a large area of lesser functional stability?

The purpose of this study was to examine hip joint stiffness in-vivo, in an attempt to answer the following questions:

1) Does limited hip rotation indicate increased rotational stiffness?

2) Does excessive hip mobility result in a larger neutral zone?

7.2. Methods

Young adult males participated in this study, having met the criteria of demonstrating excessive or limited range of hip rotation. They were secured in an upright standing position with the leg of interest atop a turntable, allowing relatively frictionless rotation into internal and external rotation of the hip. A rotation moment was applied to the turntable, while hip rotation angle and applied force data were collected. Electromyographic (EMG) activity of the hip and thigh muscles were also monitored, to ensure that the rotation was achieved with minimal effect from the associated muscles.

7.2.1. Participants

Recruitment procedures and experimental methods were approved by the university human research ethics committee, and all participants completed a written informed consent document. Male participants between the ages of 19 and 30 were recruited from the university population and surrounding area via posters and word of mouth. All subjects were healthy and without current hip or back pain or past pathology in these regions. Previous research in this

Table 7-1: Percentile data of hip joint rotation from a young adult male population. 50^{th} percentile represents mean (SD).

	Percentiles: Hip rotation in prone lying						
	5th	**10th**	**25th**	**50th**	**75th**	**90th**	**95th**
Total Rotation	44°	46°	53°	59(11)°	66°	75°	82°
Internal Rotation	12°	15°	20°	26(8)°	31°	37°	42°
External Rotation	19°	23°	28°	34(9)°	40°	46°	50°

laboratory analyzed prone lying hip rotation in a group of 77 males, between the ages of 19 and 30 to determine normative and percentile data for this age group (Table 7-1). In that the purpose of this study was to compare the two extreme percentile groups, participants were sought who demonstrated either greater than 75° or less than 46° of total hip rotation, representing greater than the 90th and less than the 10th percentile. These two groups will hereafter be referred to as the excessive hip mobility (EHR) and limited hip mobility (LHR) groups, respectively. Hip rotation measurements were obtained in a prone lying position, knee flexed to 90°. The same investigator passively rotated the leg of interest, while stabilizing the pelvis. Rotation measurements were obtained by an assistant, using a standard goniometer. In total, 18 participants were recruited; the data from 2 were not used due to high electromyographic (EMG) activity and/or ankle instability problems. The average hip rotation range of motion (ROM) for each group is shown in Figure 7-2. The leg which best represented the LHR or EHR group was used for analysis.

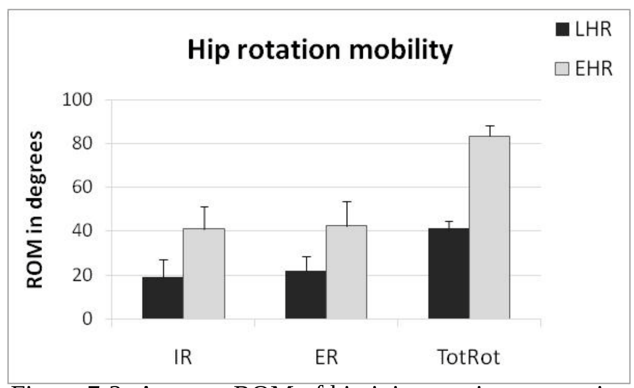


Figure 7-2: Average ROM of hip joint rotation comparing the two groups: those with LHR and those with EHR. (n = 8 per group)

7.2.2. <u>Data collection</u>

Participants stood with the feet on two platforms: the leg of interest standing on a circular board which was mounted on top of turntable apparatus with sealed bearings, resulting in relatively frictionless motion. The other leg stood on a non-mobile stool of the same height. The participant was positioned such that the centre of rotation of the platform was directly under the participant's tibial shaft, with minimal flexion/extension or ab/adduction of the hip, in an attempt to approximate a vertical line from the centre of rotation of the jig to the hip joint. The foot was secured with a Velcro™ strap to prevent slippage. Adjustable wooden phalanges on either side of the pelvis were able to slide towards each other, thus providing a secure hold to prevent lateral pelvic motion. Non-elastic straps around the thorax and pelvis further restricted thorax and pelvis motion (Figure 7-3). Participants were instructed to attempt to minimize muscle activity around the hip joint and allow the leg to be pulled into rotation: both IR and ER. They were to inform the investigator if any pain was felt due to the procedure.

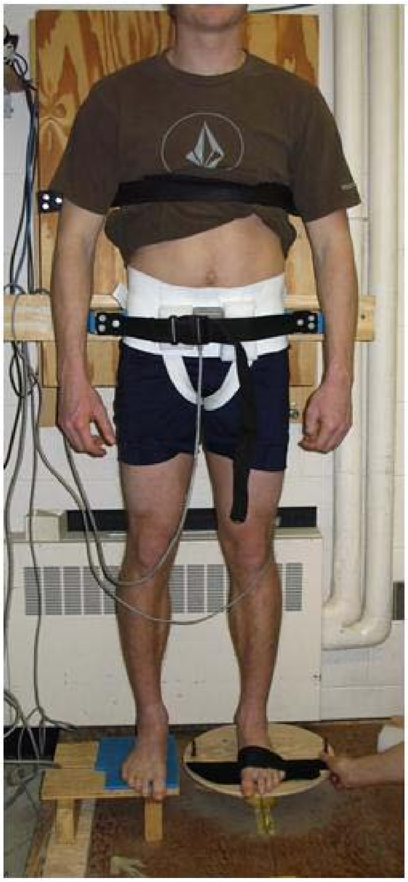


Figure 7-3: Participant with the left leg resting atop the passive rotation device.

An initial calibration posture was collected, such that the long axis of the foot was facing directly forwards, as shown in Figure 7-3. This was considered 0° of rotation. There followed a series of 8 data captures: 4 passive (2 IR, 2 ER) and 4 active (the latter to be used for a different study). The order of capture was randomized. The leg was pulled into either IR or ER, using 2 eye-bolts and a carabineer in series with the force transducer (Figure 7-4). The angle of pull was maintained such that it was perpendicular to the medial/lateral diameter of the platform, and was applied until the investigator was unable to perceive any further rotation of the limb. Trials were repeated if there was noticeable buckling of the knee or ankle, or if the associated EMG appeared to be more than minimally active in the raw state. Two trials were collected for redundancy: the first trial was used for analysis unless there were EMG or artifact problems.

7.2.3. *Instrumentation*

To ensure that rotational motions were purely passive, surface EMG data was collected from the following ipsilateral hip/thigh muscles: gluteus medius, gluteus minimus, biceps femoris, and quadriceps/psoas. Pairs of Ag-AgCl surface electrodes were positioned over muscle bellies with an inter-electrode distance of 3 cm, in line with the direction of the muscle fibres. Signals were amplified (± 2.5V; AMT-8, Bortec, Calgary, Canada; bandwith 10 – 1000 Hz, common mode rejection ratio (CMRR) = 115 db at 60 Hz, input impedence = 10 GΩ), captured digitally at 2048 Hz, low pass filtered at 500 Hz, rectified and low pass filtered at 2.5 Hz (dual pass) and normalized to the maximum voltage produced during isometric maximum voluntary contraction (MVC) trials to produce a linear envelope. EMG activity was observed during data collection, and trials were repeated which appeared to have noticeable activity in response to the application of the rotation moment. Post-processing was used to ensure that EMG activity in all muscles remained lower than 5% MVC.

Figure 7-4: Left hip ER, illustrating the eye-bolts and carabineer in series with the force transducer.

Three-dimensional hip motion was recorded using an electromagnetic tracking device (Isotrak, Polhemus, Colchester, VT, USA) with the source secured over the anterior pelvis via a non-elastic strap and the sensor affixed to the lateral femoral condyle via two-sided tape and secured with a flexible Velcro™ strap. Hip motion data was sampled digitally at 32Hz and synchronized to the EMG data. All angular measurements were made relative to the standing anatomical position.

Applied forces were recorded with a force transducer (Transducer Techniques Inc., Temecula, CA, USA) and digitally sampled at 2048 Hz. These data were dual-pass filtered (second order 6Hz low pass Butterworth), and downsampled to 32 Hz to match the hip positional data.

These forces were used to calculate moment/angle curves as described below. In addition, the moment required to initiate rotational movement of greater than 1°, and the subsequent angle of rotation at 6 Nm of applied moment were also extracted from the data.

7.2.4. Moment-angle curves

The applied moment was calculated as the product of the measured force and the moment arm, (0.15m for all collections). This was plotted against the angle of rotation, which had been zero'd relative to the starting position in each trial. These plots were windowed for each trial and normalized in time to ensure equal trial lengths for each participant, each trial. Data were then combined across subjects for each direction of rotation, resulting in 4 moment/angle curves. Exponential curve fits were performed, as previously described by McGill et al (1994), resulting in 4 moment/angle curves, each represented by the following equation

$$M = \lambda e^{\delta\,\varphi}$$

Where M = applied moment (Nm)

φ = angle of rotation

λ, δ = curve-fitting coefficients

This curve was further differentiated with respect to angle, resulting in hip angular stiffness (K in Nm/degree of rotation) where:

$$K = \delta\lambda e^{\varphi\,\delta}$$

7.2.5. Variability

As mentioned previously, there were a total of 8 trials per participant (4 for another study). The angle of hip rotation prior to the application of force was documented for each participant. These angles were averaged over the 8 trials for each participant, resulting in an average starting position, and range of starting positions. These numbers were then averaged over each group (LHR/EHR). The purpose of this was to gain insight into whether any significant difference existed between the groups, as to the tendency for them to return to the exact same position (indicating a small neutral zone), or whether they had a large range of positions representing elastic equilibrium, thus indicating a larger neutral zone.

7.2.6. Statistical Analysis

All analyses utilized SPSS (version 17) with a significance level set at $p < 0.05$. Independent samples t-tests were conducted to evaluate differences between the LHR and EHR groups for initial angular position of the hip, and mean range of starting positions. Bonferroni adjustments were applied. For each of the dependant variables of: 1) moment to initiate movement and 2) angle of rotation at 6 Nm of applied moment, a repeated measures ANOVA was conducted, with IR and ER as the within subject factors and LHR/EHR being the between subject factors. Post-hoc tests were conducted using SAS (version 9.2).

7.3. Results

7.3.1. *Starting position and Neutral zone*

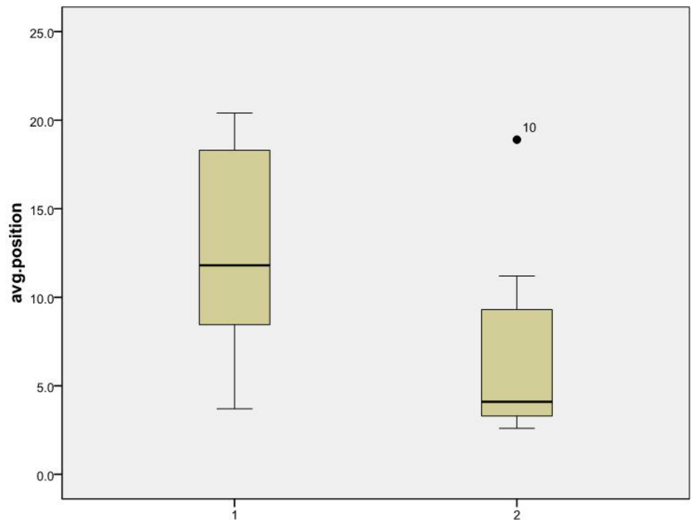


Figure 7-5: Average angle of hip ER prior to the application of rotational force in the LHR group (1) compared to the EHR group (2). These results were averaged over 8 trials for each participant, then the average of each group was calculated, and demonstrate the increased angle of ER adopted by the group with LHR. (n = 8 per group)

As a group, the men with limited hip rotation tended to adopt an average standing posture with 4.9° more external rotation of the hip, compared to the EHR group (p = 0.176) (Figure 7-5). The statistical power of this test was 0.39; thus, increasing the number of participants to 11 per group would have resulted in attaining significance. Despite this increase in standing ER angle, hip restriction patterns in prone lying was almost identical when comparing with the EHR group, for both IR and ER (Figure 7-2). The LHR group also demonstrated a greater range of starting positions, prior to force application. That is, over the course of 8 randomly ordered IR and ER trials, the position of elastic equilibrium prior to the application of force varied over 11.6(6.1)° in the LHR group compared to

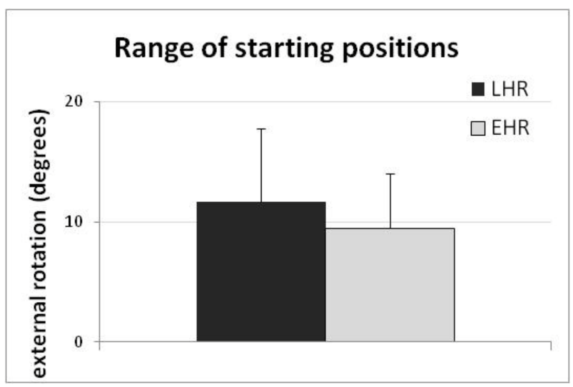


Figure 7-6: Average range the initial positions adopted prior to the application of a rotational moment. These results are averaged over 8 trials for each participant, then averaged over all participants. (n = 8 per group)

9.4(4.6)° in the EHR group (Figure 7-6). Similarly, the initial "toe region" of the moment/angle curves did not differ greatly between the two groups, as shown in Figure 7-7. This may have been affected by the fact that participants were weight-bearing on the leg being rotated. Consequently, a reasonable torque was required to initiate and increase rotation. This differs from previous research analyzing passive stiffness on a frictionless jig (Beach, Parkinson, Stothart, & Callaghan, 2007; Brown & McGill, 2008, Parkinson, Beach, & Callaghan, 2004; Scannel & McGill, 2003), in which the participants were non-weight bearing (NWB). Likely, in a NWB environment, after movement is initiated, little additional torque would be required to continue angular motion until passive tissue resistance is met. In this current study, however, the weight of the upper body and leg would add to the friction/compression in the hip joint, thus reducing its ability to rotate freely.

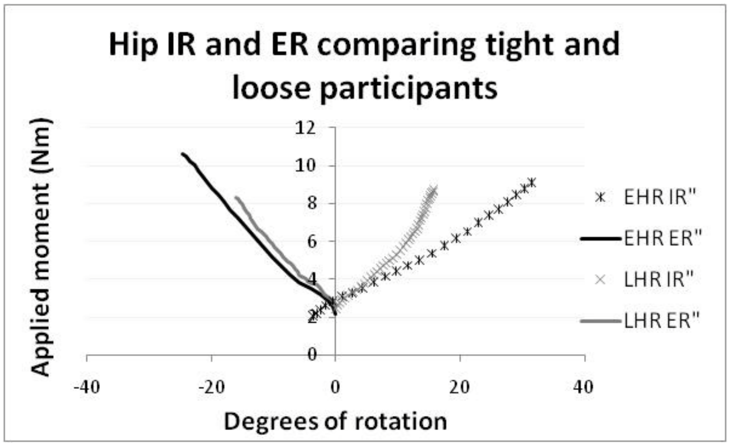


Figure 7-7: Passive hip IR and ER moment/angle curves comparing 2 participants: one from each of the hip mobility groups. This graph demonstrates that the difference in mobility in the hips may be due more so to tissue tension differences throughout the range of motion than specific disparities in the neutral zone. (n = 8 per group)

7.3.2. <u>*Force / movement*</u>

Significantly more applied torque was required to initiate rotation in the LHM hip group compared to the EHR, when collapsed across both IR and ER ($p = 0.038$)

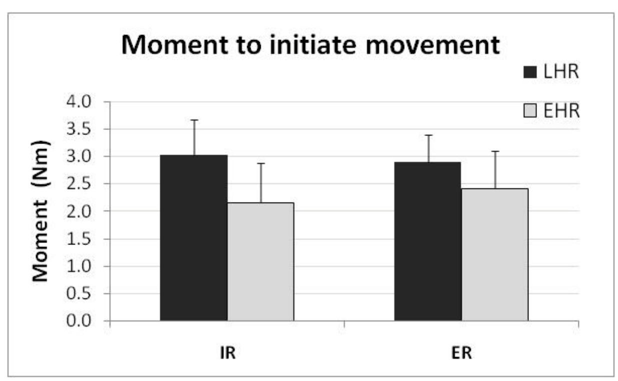


Figure 7-8: Rotational moment required to induce hip rotation of greater than 1 degree. (n = 8 per group)

(Figure 7-8). LSD post-hoc tests resulted in a significantly greater moment required by the LHR group in the IR condition (p = 0.024), but not ER (p = 0.164). Specifically, the LHM group required 0.84 Nm and 0.44 Nm more than the EHR for IR and ER, respectively. This translates into force differences between LHM and EHR of 5.4 N and 2.8 N (IR and ER) due to the small moment arm (0.155 cm) from the application of force to the centre of rotation (force = torque / moment arm). At a moment of 6 Nm, which was approximately the mid-point of most of the trials, those participants with EHR hips had rotated significantly further than those with LHR when collapsed across both directions (p = 0.004) (Figure 7-9). LSD post-hoc testing determined a significantly greater angle of rotation in the EHR group in external rotation (p = 0.0012), but not IR (p = 0.1008).

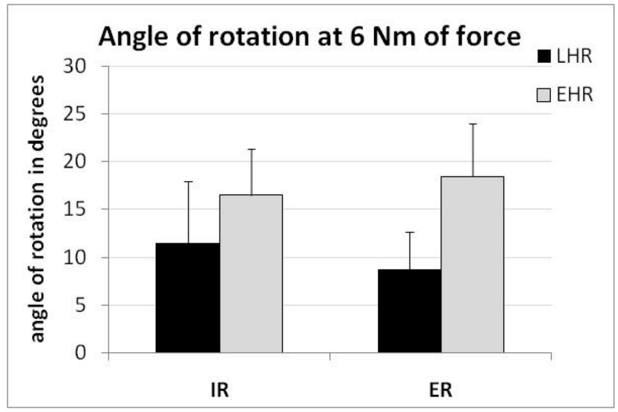


Figure 7-9: Angle of hip rotation at the point in time when 6 Nm of rotational moment was applied. (n = 8 per group)

7.3.3. *Stiffness curves*

Exponential curve fitting was conducted on the stiffness (moment/angle) curves (Figure 7-10), with resulting formulas for each of the curve fits listed in Table 7-2. Each plot is cumulative for all participants in that group, with

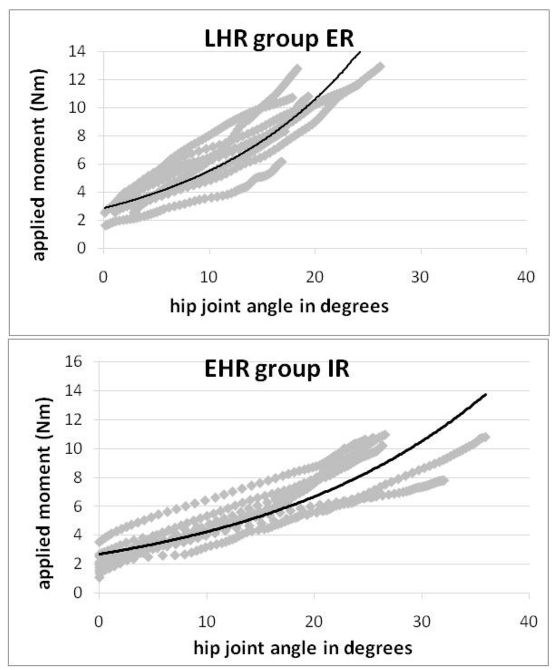


Figure 7-10: An example of the moment/angle scatterplots with exponential curve fits. The data includes all participants in the respective LHR or EHR groups for the rotational direction specified. (n = 8 per group)

separate graphs for IR and ER, LHR and EHR. The derivative of these exponential curve fits was then plotted against 0° – 35° of rotation (Figure 7-11) (Brown & McGill, 2008). Stiffness increased exponentially in all conditions, with the greatest increase over angle occurring in the LHR/ ER condition, which was approximately 2.5X stiffer than the least stiff group (EHR/ER) at 25° of rotation (the maximum angle attained in the LHM/ER group). Stiffness in the EHR/ER and LHM/IR group appear almost identical.

Table 7-2: Formulas for the exponential line of best fit for the two groups, two rotational directions, as well as the associated R^2 values. (n = 8 per group)

	Exponential curve fit	R^2 value (Nm)
LHR/IR	$y = 3.6096e^{0.0439x}$	0.523
LHR/ER	$y = 2.8622e^{0.0655x}$	0.774
EHR/IR	$y = 2.6986e^{0.0454x}$	0.777
EHR/ER	$y = 2.0412e^{0.0541x}$	0.735

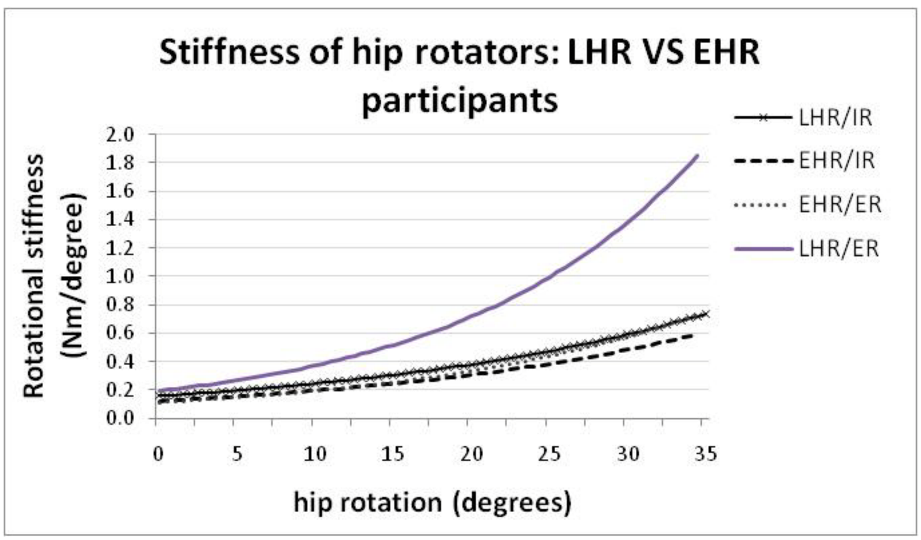


Figure 7-11: Stiffness of the hip joint in rotation, determined from the first derivative of the moment/angle curves from **Error! Reference source not found.**, across a ROM of 35 degrees. (n = 8 per group)

7.4. Discussion

Young adult males with limited hip mobility tend to stand with more external rotation of the hip, potentially pre-tensioning the strong ilio-femoral ligament: a major stabilizer of the anterior hip joint, while decreasing the tension being applied to the posterior joint structures.

Further passive ER of the joint in the LHR group demonstrates much greater stiffness than does IR, or either motion in the EHR group, possibly due to this pre-tensioning. This position may also explain why the LHR group did not exhibit greater stiffness in the IR direction compared to the EHR group, despite significantly less ROM available in the pre-screening. These findings are in keeping with previous in-vitro investigations of the hip joint, which describe significantly lower failure strain and greater stiffness in the anterior ilio-femoral ligaments, compared to the posteriorly located ischiofemoral (J. Hewitt et al., 2001; J. D. Hewitt et al., 2002). An ER torque applied to the hip has been shown to cause femoral head displacement in an anterio-inferior direction, and has been suggested as one of the mechanisms for labral tears in a young athletic population (Crawford et al., 2007; Dy et al., 2008; McCarthy et al., 2003).Thus, a group with limited ER and increased stiffness in these structures may be more at risk for injury, especially when the rotational torque through full range occurs rapidly, thus less ligament viscosity (Bigliani et al., 1992; Latash & Zatsiorsky, 1993). To the best of our knowledge, there are no comparative studies on passive hip joint rotational stiffness in-vivo. While it is not possible at this time to differentiate ligamentous from muscular tension in-vivo, the internal rotators of the hip are not a physically large group in cross-sectional area, and the ilio-femoral ligaments are generally credited with being the main structure limiting ER (Crawford et al., 2007; Gray, 1974).

When averaged over each group, there was no evidence that the neutral zone was larger in the group of males with EHR hips compared to the LHR group. While it was anticipated that those with excessive hip rotation would show a greater range of starting positions, thus more variability, this was not the case. This concept might, in theory, be better analyzed in a non-weight-bearing (NWB) situation, as has been done in the lumbar spine, allowing the hip to freely rotate within the neutral zone with minimal applied torque, (Beach, Parkinson, Stothart, & Callaghan, 2007; Brown & McGill, 2008; Parkinson, Beach, & Callaghan, 2004; Scannell & McGill, 2003; Scholtes et al., 2009). However, applying a rotational torque to a NWB leg would have involved a different equipment setup, and the investigators decided that studying a weight-bearing hip joint had greater functional applications. In addition, to impart a rotational torque to a NWB leg would lessen compression in the ankle and knee, perhaps allowing more torque absorption in these joints, and confounding the results. To apply a torque directly via rotation of the thigh would be difficult to quantify due to associated soft tissue movement.

Limitations exist, in that these data reflect stiffness only in a select group of young male adults. This was done to limit the variability that might occur between sexes as well as the possibility of changes in ligamentous laxity over the menstrual cycle which has been described in females (Chandrashekar et al., 2005; Chandrashekar et al., 2006; J. Slauterbeck et al., 1999; J. R. Slauterbeck et al., 2002; Wojtys et al., 2002). Assumptions were made that the lower limb was a vertical cylinder from the centre of the rotation platform to the hip joint, and that minimal rotational torque would be absorbed by the ankle and knee joints. Every attempt was made to ensure that the participants' hip being rotated was in minimal flexion/extension or ab/adduction, and that the foot was placed such that the centre of rotation was under the shaft of the tibia. The possibility also exists, that those participants with laxity in the hips would also demonstrate laxity in the knee and ankle joints, which may have affected the outcomes of hip stiffness. This concept is worthy of further investigation. Finally, a larger sample population might have lent more statistical power to the analyses. Studying people who fit into extreme groups, while maximizing the difference between groups, also limits the number of participants who qualify for participation.

7.5. Summary

In a group of young adult males with limited hip mobility, mechanical stiffness was much greater in the anterior hip joint structures than that demonstrated in a group with excessive mobility. Consequently, extreme rotational torque may result in soft tissue tearing, as opposed to stretching and absorption of the tension, as would be better accomplished in looser, less stiff hips. The results of this study do not indicate that young men with excessive hip mobility have a larger neutral zone, thus no evidence that their hip is clinically less stable. Consequently, it does not appear that clinicians need focus on stabilizing hip joints with excessive mobility: the increased ROM is more so due to overall stiffness changes in the soft tissues constraining motion (as would be depicted by scenario “b” in Figure 7-1), and not because of an overly large neutral zone. Thus, on average, it appears that increasing flexibility in the LHR group would do more to prevent injury than increasing stability and/or decreasing mobility in the EHR group. Further investigation is warranted into changes that occur in soft tissue stiffness over the course of a flexibility program: is it stiffness, or neutral zone that is gained? Similarly, looking at differences between LHR and EHR groups in joints other than the hip would help elucidate the ideal balance between flexibility and stiffness of soft tissue structures surrounding and supporting the joints of the human body.

8. Limited hip extension: its effect on passive joint stiffness and the hip/back extension relationship

8.1. Introduction

Little is known about passive resistance to hip extension in-vivo. Since it is relatively easy to stabilize the pelvis while rotating the hip around a longitudinal axis (hip internal rotation (IR) and external rotation (ER)), numerous studies have analyzed the effect of passive hip rotation on the spine and low back pain in non-weight bearing. Results indicate that a low back pain population tends to have less hip rotation and more asymmetry of hip rotation (Chesworth et al., 1994; Cibulka et al., 1998; Van Dillen, Gombatto, Collins, Engsberg, & Sahrmann, 2007; Van Dillen et al., 2008) . Similarly, those with low back pain tend to display earlier lumbo-pelvic motion in association with active prone lying hip rotation (Scholtes et al., 2009; Van Dillen et al., 2007). However, despite knowing that lack of hip joint extension affects the lumbar spine (Kujala et al., 1992; Kujala et al., 1994; L. W. Lee et al., 1997; Mellin, 1988; Mellin, 1990; Offierski & Macnab, 1983; Riley et al., 2001; Schache et al., 2000; Van Dillen et al., 2000), there appears to be little information regarding the effect of limited hip extension on the lumbar spine in a passive or non-weight bearing environment. This information would give insight into the innate stiffness properties of the hip and back, and identify regions of least resistance to an extensor moment in the hip/back complex. Comparing two groups of young adults whose hip extension mobility fits into the top and bottom percentiles of a normal distribution may give insight into the differences that occur between these two extremes, assisting clinicians and rehabilitation workers to optimize treatment protocols for the hip and low back.

Hip joint extension is primarily limited by the ilio-femoral, or "Y-ligament of Bigelow" (Gray, 1974), which is intimately attached to the superficial aspect of the joint capsule. From their origin on the anterior inferior iliac spine (AIIS) and acetabular rim, the superior and inferior aspects of the iliofemoral ligament crosses over the anterior aspect of the joint to insert along the intertrochanteric line on the femur (Gray, 1974; J. Hewitt et al., 2001). This pair of ligaments has been shown to withstand higher tensile force with less material strain at failure than other hip ligaments (J. D. Hewitt et al., 2002), providing a strong passive restraint to hip extension, thus allowing erect posture to be maintained without active muscular control. Specifically, the inferior ilio-femoral ligament resists significantly more of an extension moment than other hip ligaments (Hidaka et al., 2009). Active resistance to hip extension is mainly provided by the psoas, iliacus and rectus femoris (RF) muscles. Psoas originates from the T12 – L5 vertebrae, and has been shown to stabilize the lumbar spine in sitting, or when a heavy load is applied on the contralateral side (Andersson, Oddsson, Grundstrom, & Thorstensson, 1995). Iliacus begins in the anterior aspect of the ilium, to conjointly insert on the lesser trochanter with psoas. The main action of both of these muscles, however, is hip flexion (Andersson et al., 1995; Andersson, Oddsson, Grundstrom, Nilsson, & Thorstensson, 1996; Juker, McGill, Kropf, & Steffen, 1998). The RF, on the other hand, from its origin on the AIIS, crosses both the hip and knee joints, thus is considered a "two joint" muscle. While it is known to assist hip flexion, its primary action is thought to be knee extension, in combination with the other muscles that make up the quadriceps group (Kendall & McCreary, 1983).

Most studies analyzing the effect of limited hip extension on the lumbar spine have been conducted on the elderly or neurologically impaired, in which case the etiologic nature of their hip impairment may also cause abnormal biomechanics of the lumbar spine. In any set of normative data, there exist those people who are in the extremes of the normal distribution, thus presenting with excessive or limited range of motion (ROM). Accessing such groups in a young adult population may lend insight into the effect that limited hip extension has on the lumbar spine, while minimizing the likelihood that lumbar motion will be affected by arthritic changes or altered neural control.

The purpose of this study was to compare two groups of men: those with excessive hip extension (EHE) and those with limited amounts of hip extension (LHE). The hypotheses being tested were:

1. Men with LHE will adopt a position of elastic equilibrium which is more flexed relative to the pelvis, compared to a similar group with EHE.
2. When an extension moment is passively applied to the lower leg, those with LHE will demonstrate greater lumbar spine extension than those with EHE.
3. Hips with limited extension will demonstrate increased stiffness in response to an applied extension moment, compared to hips with excessive extension.

8.2. Methods

A total of 18 healthy men, representative of those with LHE and EHE, took part in the study. They were secured in left side lying in an apparatus which isolated sagittal motion to the left hip, the lumbar spine, or a combination of both. Motion was relatively frictionless, as the apparatus sections supporting the pelvis and lower extremity were atop a series of nylon precision bearings. The torso and upper body were secured in an immobile section. An extensor moment was applied to the distal segment. Motion at the hip and spine were monitored with an electromagnetic sensor. Electromyographic (EMG) activity of the torso, thigh and gluteal muscles was monitored and collected, thus ensuring all motions were passive.

8.2.1. Participants

Recruitment procedures and experimental methods were approved by the university human research ethics committee, and all participants completed a written informed consent document.

Male participants between the ages of 19 and 30 were recruited from the university population and surrounding area via posters and word of mouth. All subjects were healthy without current hip or back pain or past pathology in these regions. Previous research in this laboratory analyzed supine hip extension in a group of 77 males, between the ages of 19 and 30 to determine normative and percentile data for this age group. Results are shown in Table 8-1. In that the purpose of this study was to compare LHE and EHE groups, participants were

Table 8-1: Percentile data representative of a normal distribution for hip extension. 50^{th} percentile represents the mean ± standard deviation.

	Percentiles: Right/Left hip extension, collapsed						
Percentile	5th	10th	25th	50th	75th	90th	95th
Modified Thomas Test (MTT)	-18°	-16°	-12°	-8(6)°	-5°	-2°	4°
Relative to horizontal, no correction for pelvis	+8°	+6°	+2°	-2(6)°	-5°	-8°	-14°

sought whose hip extension measurement fit into either the 0 - 20^{th} percentile or the 50 – 100^{th} percentile. These two groups will hereafter be referred to as the LHE and EHE groups, respectively.

Hip extension measurements were conducted using the modified Thomas test (MTT) (Boone et al., 1978; Gabbe et al., 2004; Kendall & McCreary, 1983) with the investigator controlling for ab/adduction and rotation. As depicted in Table 8-1, the MTT calculates hip extension relative to a pelvis flexed approximately 10°. Despite the fact that this number better represents the true hip/pelvis angle, the research team chose to label hip extension relative to the horizontal for the purpose of participant categorization. This number is listed below the MTT results in Table 8-1, and is how the angles were recorded during participant screening. To assist with control of the pelvis position, a blood pressure (BP) cuff was placed under the lumbar spine, and was inflated to 60 mm Hg. This pressure was monitored as one of the participant's legs was lowered passively to a position of maximum hip extension. The opposite leg was held passively in a position of hip/knee flexion by the investigator, which maintained the pressure in the cuff by preventing lumbar extension. Participants were encouraged to give feedback as to their perception of pelvis position, in an attempt to further minimize pelvic rotation during hip extension. The position of the extended thigh was measured by an assistant with a standard 360° goniometer with 2 spirit levels, one on each of the arms, to improve accuracy of determining horizontal and vertical positioning (Gabbe et al., 2004). All measurements were recorded as the thigh angle relative to the horizontal axis. A total of 21 participants were recruited for the study. After data collection and processing, the data from 3 participants could not be used as their EMG activity was higher than acceptable for a passive study (>5% MVC). Thus, each group included the data from 9 participants. Average hip extension ROM was 12.2(4)° and -6.0(4)° for the LHE and EHE group, respectively (Figure 8-1). Individual measurements are provided in Appendix 8-1.

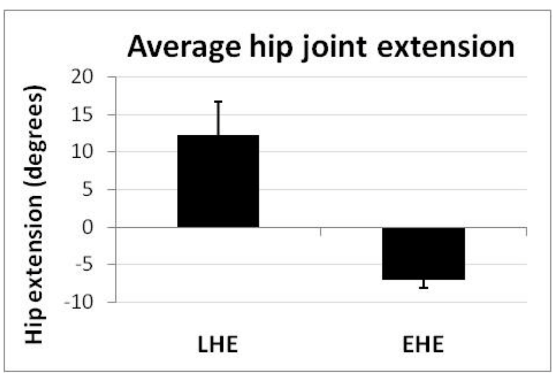


Figure 8-1: Average hip extension of the two participant groups. Measurements were made using the MTT for positioning, but represent thigh angle relative to the horizontal. (n = 9 per group)

8.2.2. Data collection

Participants lay on their left side on an immobile upper body platform, with non-elastic straps securing their upper torso to the platform and vertical uprights that were anterior to their chest. Their pelvis was secured to a small independent platform with a vertical post, while their legs were secured to a third platform; these two mobile platforms had a plexiglass bottom surface and were free to glide over a similar surface of plexiglass with precision nylon bearings between the surfaces, resulting in a relatively frictionless motion (Figure 8-2). The three platforms were attached to each other via a series of eye-bolts and turnbuckles (Figure 8-3).


Figure 8-2: Position of participant on the frictionless jig. In this specific scenario, the right leg is suspended from the ceiling, allowing free movement of the left hip.

Selective removal of the turnbuckles would allow motion to occur either between the thorax and pelvis, the pelvis and thigh, or a combination of the two (Figure 8-4). Some of the trials required the right leg to be secured atop the left leg, thus both would move as a unit, while for others the right leg was suspended from the ceiling via a

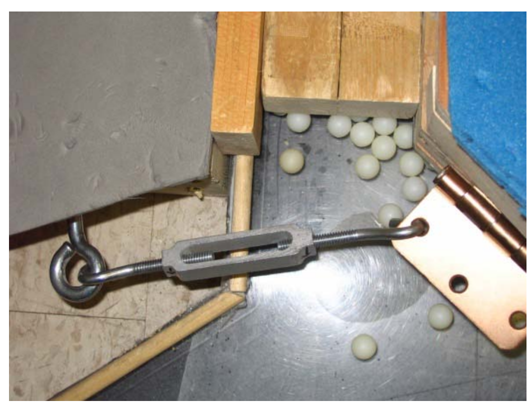


Figure 8-3: Close-up view of the turnbuckle and bolt/hinge system used to secure the upper and mid-sections of the frictionless jig.

sling apparatus, to allow free motion of the left leg only (as shown in Figure 8-2). During the entire collection, every attempt was made to ensure the participant's comfort by using of padding, and encouraging them to inform us of pressure points or general discomfort.

Participants were taken through a series of passive motions which were applied in random order. They were encouraged to indicate via a tapping cue if they were in unreasonable discomfort or wished the trial to end. All trials were repeated a minimum of two times.

1. Left hip extension with the knee straight: The right leg was suspended from the ceiling, pelvis secured (assistant applied overpressure to ensure minimal pelvis motion), then the hip was passively pulled into maximum extension until no further motion could be detected.
2. Left hip extension with the knee bent and secured at 90°: As with #1, except the left knee was bent to 90° and secured in that position. Elastic equilibrium of the hip was allowed to re-stabilize prior to initiation of each extensor moment.
3. Concurrent back and hip extension: All turnbuckles were removed, resulting in free motion in the lumbar spine and hip. These trials were repeated with the right leg in both positions: suspended from the ceiling as well as secured to the left leg. While the extensor moment was applied until end range was perceived by either the investigator or the participant, there were many trials where joint end range was not reached due to abutment of portions of the frictionless jig, or discomfort of the participant. The data from these trials were used to observe relative amounts of back extension when the hip was in 5° and 10° of extension. All measurements were with respect to the resting position at the beginning of each trial.

The extension moment in each trial was applied to the distal end of the leg portion of the frictionless jig. The angle of pull was maintained such that it was perpendicular to the perceived line from axis of rotation (either L4, L5 or hip joint) to the distal end of the jig. Of equal interest was the position that the participants adopted prior to each extension moment, thus care was taken that they returned to a position of elastic equilibrium between trials. Photos depicting the various positions are shown in Figure 8-4.

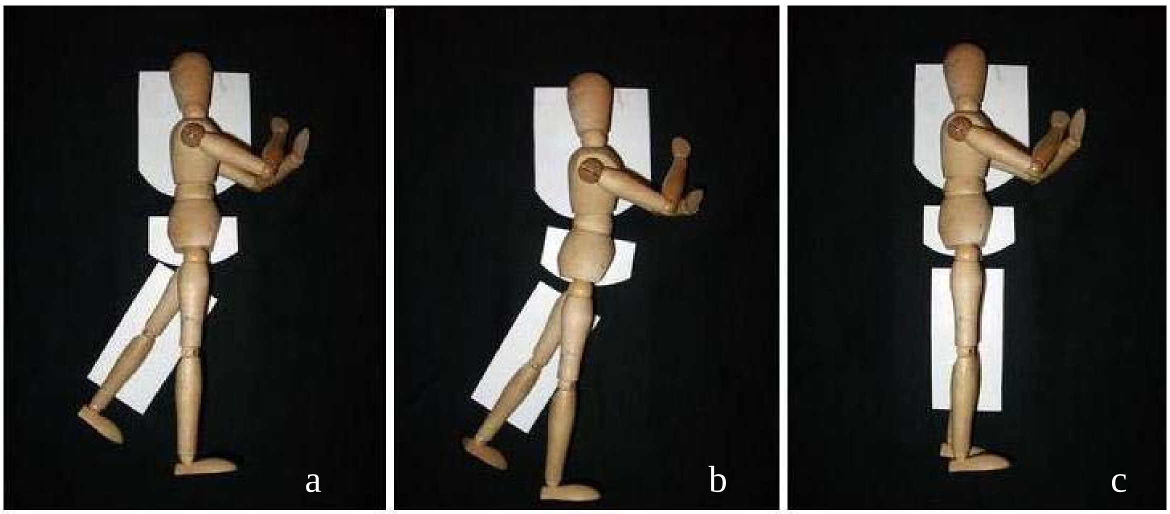


Figure 8-4: Depiction of 3positions used in this study: a) left hip extension, knee extended. b) hip and back extension. c) neutral jig position, with all portions being secured and immobile prior to segmental release.

8.2.3. <u>Instrumentation</u>

To ensure that extension motions were purely passive, surface EMG data was collected from 9 muscles: bilaterally over erector spinae (ES), rectus abdominis (RA), internal oblique (IO), as well as the left RF, biceps femoris (HS) and gluteus maximus (GM). Pairs of Ag-AgCl surface electrodes were positioned over muscle bellies with an inter-electrode distance of 3 cm, in line with the direction of the muscle fibres. Signals were amplified (± 2.5V; AMT-8, Bortec, Calgary, Canada; bandwith 10 – 1000 Hz, common mode rejection ratio (CMRR) = 115 db at 60 Hz, input impedence = 10 GΩ), captured digitally at 2048 Hz, low pass filtered at 500 Hz, rectified and low pass filtered at 2.5 Hz (dual pass) and normalized to the maximum voltage produced during isometric maximum voluntary contraction (MVC) trials to produce a linear envelope. EMG activity was observed during data collection, and trials were repeated which appeared to have noticeable activity during the application of the extensor moment. Post-processing was used to ensure that EMG activity in all muscles remained lower than 5% MVC.

Three-dimensional hip and spine motion was recorded using an electromagnetic tracking device (Isotrak, Polhemus, Colchester, VT, USA) with the source secured over the posterior sacrum via a firm elastic strap. The two sensors were affixed to the distal posterior thigh and over the T12 spinous process via two-sided tape and secured with a flexible Velcro™ strap. This motion data was sampled digitally at 32Hz and synchronized to the EMG data. All angular measurements were made relative to the standing anatomical position. These data were scrutinized in the post-processing to ensure that the range of motion had not been erroneously limited by the wooden platform components abutting against each other. Such trials were removed from the processing.

Applied forces were recorded with a force transducer (Transducer Techniques Inc., Temecula, CA, USA) and digitally sampled at 2048 Hz. These data were dual-pass filtered (second order 6Hz low pass Butterworth), and downsampled to 32 Hz to match the hip positional data. These forces were used to calculate moment/angle curves as described below.

8.2.4. Moment-angle curves

To calculate hip joint stiffness, the applied moment was calculated as the product of the measured force and the moment arm, (distance from applied force to the level of the hip joint). This was plotted against the angle of rotation, which had been zero'd relative to the starting position in each trial. These plots were windowed for each trial and normalized in time to ensure equal trial lengths for each participant, each trial. Data were then combined across subjects for each condition (LHE vs. EHE, knee flexed or extended), resulting in 4 moment/angle curves. Exponential curve fits were performed, as previously described by McGill et al (1994), resulting in 4 moment/angle curves, each represented by the following equation

$$M = \lambda e^{\delta\,\varphi}$$

Where M = applied moment (Nm)

φ = angle of rotation

λ, δ = curve-fitting coefficients

This curve was further differentiated with respect to angle, resulting in hip angular stiffness (K in Nm/degree of rotation) where:

$$K = \delta\lambda e^{\varphi\,\delta}$$

8.2.5. <u>*Statistical analysis*</u>

All analyses utilized SPSS (version 17) with a significance level set at $p < 0.05$. Independent samples t-tests were conducted to compare the initial hip extension ROM available in the participants. To analyze the initial resting position of the hip in the frictionless jig, a one-way repeated measures ANOVA was conducted, with hip mobility as a between subject factor. The relationship between passive hip extension and concurrent back extension was analyzed using a 2 way repeated measures ANOVA, with degrees of hip extension and leg position being within subject factors, and hip mobility as the between subject factor. Independent samples t-tests with a Bonferroni adjustment were used to parse out specific differences between the LHE and EHE groups for any specific interactions.

8.3. Results

As depicted in Figure 8-1 and Appendix 8-1, there was a significant difference in hip extension ROM between the two groups ($p < 0.001$) at the time of intake. When lying in the frictionless jig with available hip motion, the resting position for the hip was one of significantly more flexion when the knee was bent to 90° (Figure 8-5) ($p = 0.002$), but there was no significant difference between the LHE and EHE groups ($p = 0.251$). The EHE group, in fact, adopted a position of greater hip flexion when the knee was flexed, resulting in an average of 25.9° compared to 20.2° in the LHE group. This is probably indicative of a greater effect of rectus femoris tension in the EHE group than the LHE.

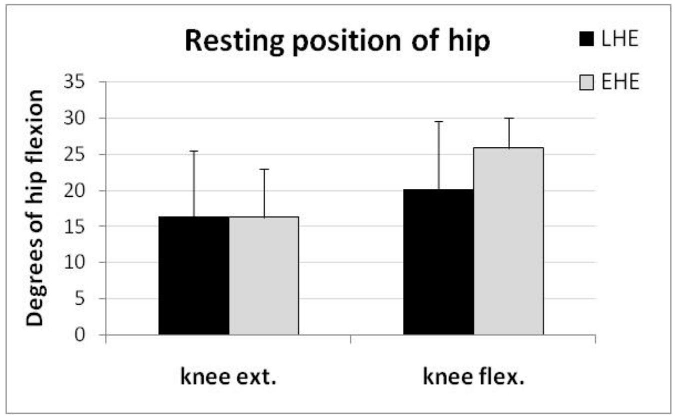


Figure 8-5: Position of elastic equilibrium for the left hip (right hip suspended), comparing the LHE and EHE groups in two conditions: knee extended and knee flexed to 90°. (n = 9 for LHE/ext, 8 for EHE/ext, and 7 for LHE AND EHE/flexed knee due to sensor movement)

When the frictionless jig was released, such that motion occurred in both the hip and back, a passive extension moment resulted in varying amounts of motion occurring at each of the joints (Figure 8-6). As would be expected, there was a significant effect for degrees of extension and leg

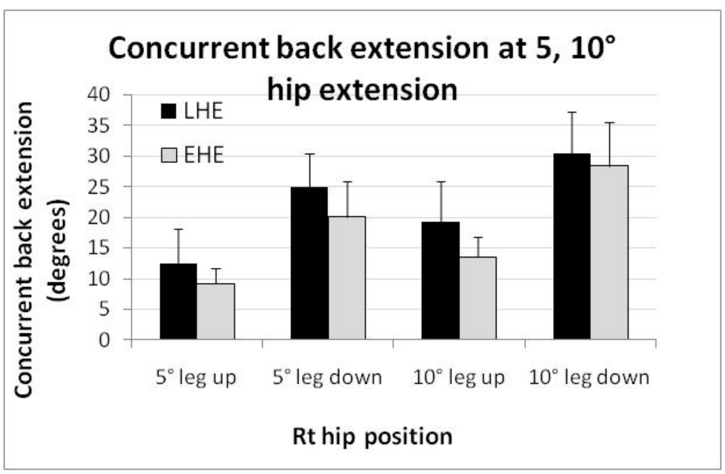


Figure 8-6: Concurrent back extension measurements when the hip had reached 5° and 10° of extension. See text for description of significance levels. (n = 9 for each group, except 8 for each of the 10° extension trials due to platform abutment)

position ($p < 0.001$ for both), with more back extension occurring when the right leg was secured to the left, as well as at 10° of hip extension, compared to 5°. The difference between the LHE and EHE group demonstrated a significance level of $p = 0.06$, thus not statistically significant. However, the statistical power of this analysis was only 0.481, and would likely have resulted in greater significance had more participants been included in the study. As can be seen in Figure 8-6, an extension moment applied to the hip and back concurrently resulted in more back than hip extension in all conditions, indicating less resistance to movement in the spine compared to the hip. Specifically, the LHE group consistently demonstrated a tendency to extend the spine to a greater degree than did the EHE group.

There was also a significant 3-way interaction ($p = 0.013$) between leg position*degrees of extension*hip mobility. Using independent t-tests with Bonferroni adjustment, the difference between LHE and EHE neared significance at $p = 0.041$ when the left hip was extended to 10° and right leg elevated; *p*-values in all other pairs were greater than 0.098.

For the hip joint, exponential curve fitting was conducted on the moment/angle curves, with resulting formulas for each of the curve fits listed in Table 8-2.

Table 8-2: Equations representing best fit exponential curves for the moment/angle curves in both the knee extended and knee flexed positions, and associated R^2 values.

	LHE Knee Ext	LHE Knee Fl	EHE Knee Ext	EHE Knee Fl
Curve fitting equation	$y = 5.4821e^{0.0528x}$	$y = 5.694e^{0.0509x}$	$y = 6.6397e^{0.0464x}$	$y = 8.5076e^{0.0419x}$
R^2	0.6166	0.6716	0.8647	0.7615

Each plot was cumulative for all participants in that group, with separate graphs for LHE and EHE groups, knee extended and knee flexed (Figure 8-7). The derivative of these exponential curve fits was then plotted against 0 - 50° of hip extension (Figure 8-8) (Brown & McGill, 2008). Despite minimal difference between the groups/conditions in the first 20° of hip extension, subsequent extension shows a trend for stiffness to increase more so in the

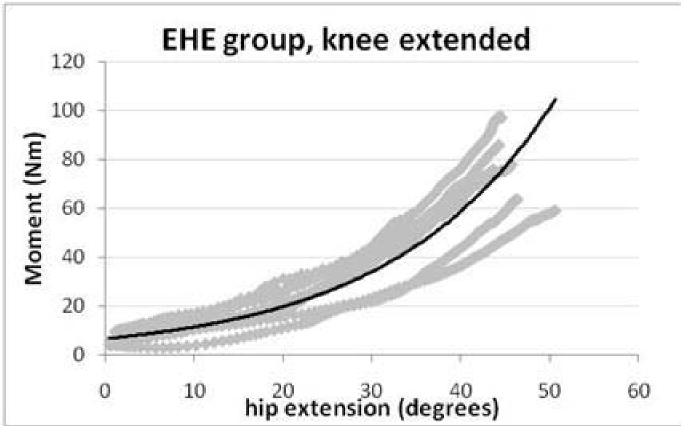


Figure 8-7: Example of a scatterplot of moment/angle data points for all participants in the EHE group, for the knee extended condition. (n = 8)

LHE group, such that at 50° of hip extension, the LHE group is approximately 25 – 30% stiffer. Specifically, in both groups, there was more resistance to hip extension with the knee in an extended position, likely due to the fact that the limiting structures would be the psoas and iliacus, in combination with the strong anterior hip ligaments. Once the knee is flexed, the tensile stress is resisted primarily by the RF. Given its anatomical structure, tensile stress would be distributed over more serial sarcomeres, potentially resulting in greater strain occurring, compared to the shorter psoas and iliacus muscles, and stiff anterior hip joint ligaments.

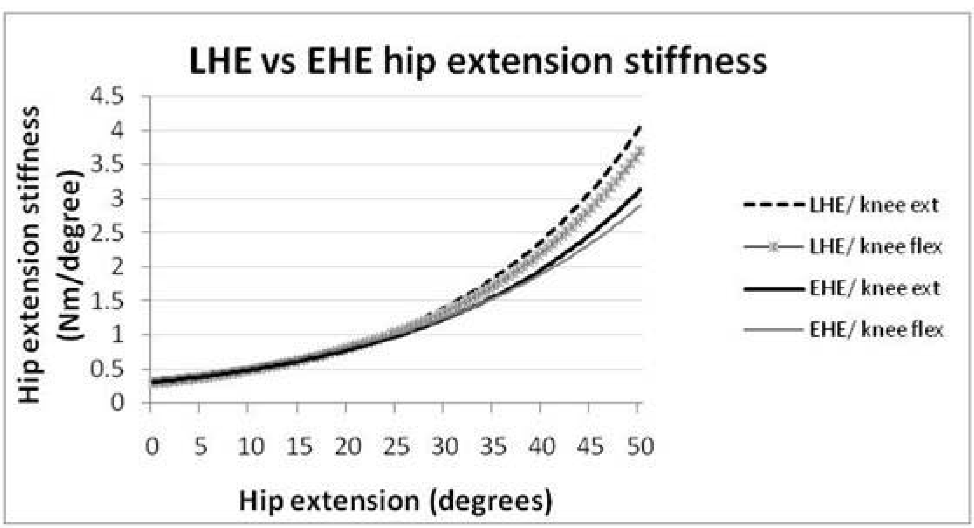


Figure 8-8: Stiffness (Nm/degree of hip extension) from the first derivative of the moment/angle curves in Table 8-2. (n = 9 for LHE/ext, n = 8 for other groups)

8.4. Discussion

With the knee extended, the position of sagittal hip elastic equilibrium in men with limited hip extension tends to be almost identical to that of men with large amounts of extension, relative to their standing posture (Figure 8-5). This brings up an interesting point which pervades all research into sagittal hip motion: how is 0° classified? Typically, as was also the case in this study, 0° represents the position adopted by each participant in upright standing. However, as was shown in an earlier study by this research team, (section 5-3) men with limited hip mobility tend to stand with greater anterior tilt of the pelvis than those with looser hips (12.4° compared to 5.4°, respectively). Consequently, the amount of hip flexion relative to the pelvis is potentially already greater in the LHE hip group at the time of the calibration pose. This infers that the true difference between the two groups in this frictionless environment might be greater than what is depicted in this study. Flexing the knee to 90° caused both groups to increase the hip flexion angle adopted in the frictionless jig, but noticeably more so in the EHE group. This might be explained in that the MTT used for screening participants did not control for knee position. Thus, for those men whose limiting structure is psoas, iliacus and anterior hip ligaments, knee position would not influence their hip extension measurement. For those with hip extension limited by RF, knee extension during this maneuver would relatively lengthen the muscle at the knee, allowing more hip extension to occur (Van Dillen et al., 2000). Consequently, it is possible that the LHE group is representative more so of men with psoas/iliacus and ligamentous tightness than RF tightness.

In a frictionless environment, with both the hip and back available to move, an extension moment causes the back to move more than the hip at a position as early as 5° of hip extension (Figure 8-6). While this may have been expected in terms of end range available ROM, this demonstrates that in the initial phase of extension the back has less resistance to movement than does the hip. Anecdotally, there exists a clinical assumption that people with limited hip extension will tend to “hinge” more in the back. That is, when actively extending the hip, they will use their back to compensate for lack of hip motion. While this study does not address active motion, it does suggest that lack of available hip extension may result in less resistance to lumbar extension: in all conditions shown in Figure 8-6 the LHE group demonstrated more lumbar extension than the EHE group in the initial 15 - 40° of total passive extension. When the right leg was suspended, and the left hip reached 10° of extension, the average difference between the LHE and EHE groups was 5.7°. Previous research in this lab conducted on men with limited and excessive hip mobility has shown that the total lumbar extension in standing averages 19° in the limited hip mobility group (unpublished data). According to Brown & McGill (2008), passive lumbar extension in a frictionless jig averages 1.5 times that obtained in standing. Thus, a difference of 5.7° between the two groups represents approximately 20% of normal ROM in this group, which is considerable, especially if the increased range tends to be focused on one or two specific intervertebral levels (Dunk et al). In addition, when both legs are secured together, in the “leg down” condition, (thus theoretically twice the hip resistance), the LHE group still demonstrated greater back extension than the EHE group, despite an overall larger amount of extension in both the LHE and EHE groups.

When an extension moment is applied to a hip joint in a frictionless environment, the greatest mid to end range stiffness is demonstrated when the knee is in extension, indicating that the stiffness of soft tissues that cross only the hip joint (psoas, iliacus, and capsule/ligaments) is higher than those that also cross the anterior knee. In addition, the group with LHE hips demonstrated more extension stiffness than the EHE group, but only in the latter half of the range (Figure 8-8). It is not possible with these data to parse out the relative amounts of musculo-tendinous vs. ligamentous/capsular resistance. Lack of electromyographic activity over the ES and RF muscle bellies indicates lack of active resistance by psoas (Juker et al., 1998), but it does not quantify the resistance attributed to their passive elements at end range.

There is little published literature with which to compare these results. Hip stiffness has generally been studied during weight-bearing activities, where motor activity was involved in creating the support moment of the limb and individual joint stiffness (Farley & Morgenroth, 1999; Hamill, Moses, & Seay, 2009; Hobara et al., 2010). Passive structures do not contribute appreciably to joint stiffness until mid-to late range, as is demonstrated by the exponential curves in **Error! Reference source not found.** Understanding the passive stiffness properties of the anterior hip joint will be a useful addition to those developing models of the hip into extension. It also provides insight into how this changes with varying knee positions, and the range of differences when comparing two groups who deviate from the average.

Limitations exist, in that these data reflect stiffness only in a select group of young male adults. This was done to limit the variability that might occur between sexes as well as the possibility of changes in ligamentous laxity over the menstrual cycle which has been described in females (Chandrashekar et al., 2005; Chandrashekar et al., 2006; J. Slauterbeck et al., 1999; J. R. Slauterbeck et al., 2002; Wojtys et al., 2002). The frictionless jig used in this study could be improved upon if future research is to rely on such an apparatus. Asking participants to side lie with the bottom leg straight resulted in undue pressure on the greater trochanter. Many participants found it uncomfortable: one stopped the data collection early, others endured in discomfort. While it would have been interesting to analyze end range back extension, with or without concurrent hip extension, attempts to obtain maximum lumbar extension frequently resulted in segments of the frictionless jig abutting against each other, obscuring true hip or lumbar end range motion. It was difficult, also, to isolate extension purely to the hip joint. Consequently, any trial which demonstrated greater than 10° of concurrent back extension was not used. This was generally noted at the time of collection, and additional trials were collected, adding manual overpressure to the pelvis to minimize lumbar motion. One cannot preclude the possibility, also, that changes may have occurred in the placement of the electromagnetic source and/or sensors during the data collection. Consequently, all movement results are presented relative to the resting position of the joint at the beginning of each trial. Finally, a larger sample population might have lent more statistical power to the analyses. Studying people whose ROM fits into the upper and lower extremes of normal give us insight into the differences that may be present. However, it also limits the number of participants who qualify for participation.

8.5. Summary

In a frictionless, side-lying environment, young adult men with limited hip extension adopted a sagittal hip position very similar to that adopted by their counterparts with excessive hip extension. Thus, despite having less total range of hip extension, their preferred position of elastic equilibrium was no different, when angles were normalized to the hip position in upright standing. The results of this study also support a clinical assumption: that people with LHE anterior hip structures will tend to use their lumbar spine as a focus for extension more so than a EHE hip group. This is important finding for those working in the rehabilitation and fitness industry, and reaffirms the importance of lumbar stabilization exercises in addition to hip flexibility stretches in such a group. Finally, the outcomes from the hip stiffness curves are a reminder that it is not enough to simply perform the modified Thomas test to determine hip extension flexibility: knee position should be standardized or tested through a range of flexion/extension to fully understand which passive structures are limiting hip extension.

8.6. Appendix

Appendix 8-1: Amount of individual hip joint extension of the participants. All measurements are made using the MTT for positioning, with numbers representative of the thigh position relative to the horizontal.

LHE		EHE	
Subj #	Hip ext	Subj #	Hip ext
2	4	3	-3
4	13	5	-12
6	15	9	-10
7	5	11	-11
8	14	13	-3
10	15	15	-4
12	15	16	-5
14	15	19	-8
18	14	20	-8
Avg	12.2		-6.0
SD	4.4		3.9

9. Comparing the elliptical trainer to walking: lumbar angles and forces, hip angles, and associated muscle activity

9.1. Introduction

The elliptical trainer has gained popularity in recent years due to its relatively low impact requirements, with a metabolic cost similar to treadmill running (Mier & Feito, 2006). The kinematics involved with elliptical use, however, are less well understood. Anecdotally, there are mixed reviews as to the effect of the elliptical trainer on the lumbar spine. While some people use it regularly with no ill effects, others claim that it provokes low back pain.

Despite its widespread use, there is little quantitative literature as to the effect of the elliptical trainer on the lumbar spine. Burnfield et al.(2010) describe an increase in lumbar flexion and corresponding decreased extension when comparing the elliptical to walking, as well as greater anterior pelvic tilt. Correspondingly, Gluteus Maximus (GMax) demonstrated higher activation levels, but no specific trunk muscle activation data were collected. Hip flexion has also shown to be increased on the elliptical, compared to level walking (Burnfield et al., 2010; Lu et al., 2007).

To the best knowledge of the authors, no literature exists which specifically address the effect of the elliptical trainer on lumbar spine kinetics and muscle activations. The purpose of this study was to analyze the effect of different hand positions, speed and stride length on hip and spine kinematics and corresponding muscle activity while using the elliptical trainer, and compare them to those demonstrated in normal walking. The hypotheses were:

1. Spine kinematics and trunk muscle activation resulting from exercising on the elliptical trainer will be the same as those found in normal walking.
2. Stride length, speed and positioning of the hands on the elliptical trainer will affect spine and hip motion as well as muscle activation.
3. Anthropometric values and hip mobility will influence joint kinematics on the elliptical trainer.
4. The elliptical trainer will cause higher phasic range of muscle activation than walking.
5. Lumbar spine forces will be greater on the elliptical trainer than those found in walking.

9.2. Methods

Healthy young men exercised on the elliptical trainer, while hand position, stride length and velocity were varied. Lumbar spine and hip joint angles were calculated, based on motion capture data collected with reflective markers and rigid linked segment modeling. Activity of the trunk and gluteal muscles was monitored with surface electromyography (EMG). Comparisons were made between the different elliptical conditions with regards to resulting hip and spine angles and resulting lumbar forces. These were also compared to those produced during normal walking.

9.2.1. Participants

Forty healthy males between the ages of 19 and 35 volunteered for this study. Their mean height was 178(7) cm and mean body mass was 79(13) kg. The participants were recruited from the University of Waterloo and surrounding area via posters and word of mouth. All claimed to be free of recent or chronic low back or hip pain or other pathology which might have interfered with participation in the study. The group included 10 graduate students, 28 undergraduates, and 2 participants who were working off-campus. Of the students, 22 were in Kinesiology, 3 in Engineering, and 13 were in various other academic departments. All of the participants except one admitted to participating regularly in some form of physical exercise. These activities included: ice hockey, soccer, track and field, basketball, football, volleyball, skiing, and general gymnasium-based fitness. Each participant completed a written informed consent document approved by the University of Waterloo Office for Research Ethics.

9.2.2. Electromyography

Surface electromyography signals were collected bilaterally on each subject from the following trunk muscles and locations: rectus abdominis (RA), 3 cm lateral to the umbilicus; external oblique (EO), approximately 15 cm lateral to the umbilicus; internal oblique (IO), halfway between the anterior superior iliac spine of the pelvis and the midline, just superior to the inguinal ligament; latissimus dorsi (LD), lateral to T9 over the muscle belly; erector spinae at T9 and L4 (T9ES and L4ES, respectively), located 5 and 1 cm lateral to each spinous process, Gluteus Maximus (GMax), located over the maximal bulk of the muscle belly, approximately mid-buttock, and Gluteus Medius (GMed), approximately 6 cm caudal to the iliac crest on the posterior-lateral pelvis. Pairs of Ag-AgCl surface electrodes were positioned with an inter-electrode distance of 3 cm. The EMG signals were collected at 2400 Hz, amplified to produce approximately ± 2.5V, and downsampled to 60 Hz to align with motion capture data. Signals were full wave rectified and low pass filtered (low pass Butterworth filter) with a cutoff frequency of 2.5 Hz, and then normalized to maximal voluntary isometric contraction (MVC) amplitudes, using a custom Labview program

(National Instruments Corp, Austin, USA). The MVC's were obtained during isometric maximal exertion tasks in the following way: for the abdominal muscles, each subject was in a sit up position and manually restrained by a research assistant, who matched the effort so that very little motion occurred. The subject produced a sequence of maximal isometric efforts in trunk flexion, right lateral bend, left lateral bend, right twist and left twist directions, but again with little motion occurring. For the extensor muscles and GMax, an isometric trunk extension was performed with the torso cantilevered over the end of the test table (Biering-Sorensen position). The MVC for GMed was measured with subjects positioned in side lying; the uppermost leg was abducted and slightly externally rotated, with a research assistant resisted maximal isometric efforts of this position. The LD MVC was performed by resisting shoulder adduction and IR at 90 degrees in the frontal plane, although in many instances the maximal activity occurred in the same MVC as the back extensors, thus was chosen for normalization. Skin preparation for the first 20 participants included shaving the electrode site and cleaning the surface with alcohol. However, there were many lost channels during collection. The last 20 participants received skin abrasion with Nuprep (Bio-Medical Instruments, Inc., Warren, USA), which resulted in minimal signal loss. Thus, the data presented for EMG amplitudes is from the latter 20 subjects.

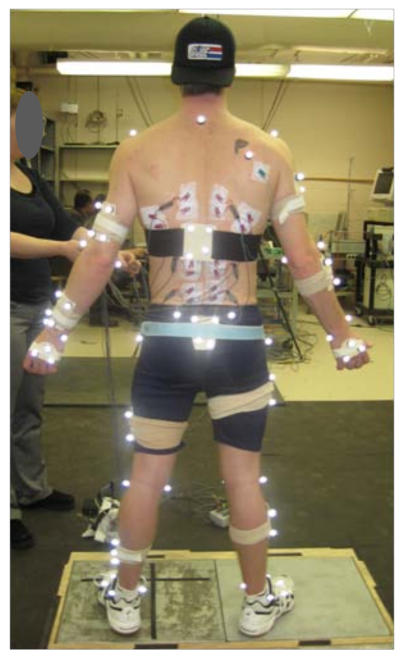


Figure 9-1: Posterior view of participant with reflective markers and EMG electrodes in place.

9.2.3. Motion Capture

A Vicon MX Motion System and Nexus software (Vicon Motion Systems, Oxford, UK) were used for capturing motion via eight infra-red cameras, collecting at a frequency of 60 Hz. Rigid plates with 4 reflective markers on each were attached via elastic straps to body segments bilaterally as follows: shin, thigh, foot, hand, forearm, upper arm, and overlying the midline on the posterior pelvis, T12 and forehead. In addition, single markers for calibration purposes only were attached over the posterior Rt scapula, C7 spinous process, sternal notch, and bilaterally over the medial and lateral aspects of each ankle, knee, wrist, elbow, ASISs, PSISs, greater trochanters, acromions, and earlobes (Figure 9-1).

9.2.4. Elliptical

An Octane (Octane Fitness, Brooklyn Park, MN USA) elliptical trainer was used for this research, as it featured variable stride lengths and was felt to represent the type of equipment commonly found in a fitness facility. Participants were invited to practice on the elliptical for as long as they felt they required prior to data collecting, to ensure basic co-ordination of movement.

9.2.5. Procedure

All testing occurred in the Human Performance Laboratory at the University of Waterloo. Participants were requested to wear spandex-type shorts, to permit application of the reflective markers on top of the shorts. T-shirts were removed, and they were asked to wear a type of footwear appropriate for running/exercising.

Anthropometric measurements were taken: height, weight, arm length, leg length, wrist to knuckle distance, as well as numerous pelvis and thorax dimensions for modeling purposes. Hip extension (using the modified Thomas test) and prone hip rotation ROM was also measured with a custom goniometer affixed with a spirit level on each arm. Following MVC collection and application of reflective markers, a calibration posture was captured with the participant standing in anatomical position. Calibration markers were then removed. Motion capture began with the participants being asked to walk at a comfortable pace along the length of the laboratory. A force plate was not used, as pilot testing had indicated that constraining the foot position to a specific target resulted in abnormal walking patterns (stride length and cadence changes). Resulting walking speeds, therefore, ranged from 41 – 60 cycles per minute (cpm), with the mean being 51.1(4) cpm. Stride length ranged from 56 – 90 cm, with a mean of 73.7(8) cm (22 – 36 inches, mean 29(3) inches, for comparison with elliptical specifications). Active lumbar ROM was also collected for flexion, extension and rotation. All trials were repeated twice.

Participants were then asked to begin exercising on the elliptical at a self-selected speed which they would choose if expecting to exercise for 30 minutes. This speed varied between 40 and 70 cycles per minute, with a mean speed of 53(7). They were encouraged to practice all 3 hand positions: holding onto the moving handles, holding onto a stationary central bar, or not holding on at all (freehand) (Figure 9-2). Besides varying hand position, stride length was varied from 18” to 26”, which were the two extremes of the Octane elliptical trainer, and speed was varied from the self-selected speed to one that was 30% faster. Once ready, data collecting moved smoothly from one position to another without stopping in between, although participants were encouraged to alert us if they felt they were getting fatigued and then were allowed to rest until they felt ready to return to exercise. Order of collection was randomized for speed, stride length and hand position. Two collections were obtained for each combination of variables, with approximately 4 cycles of elliptical motion in each.

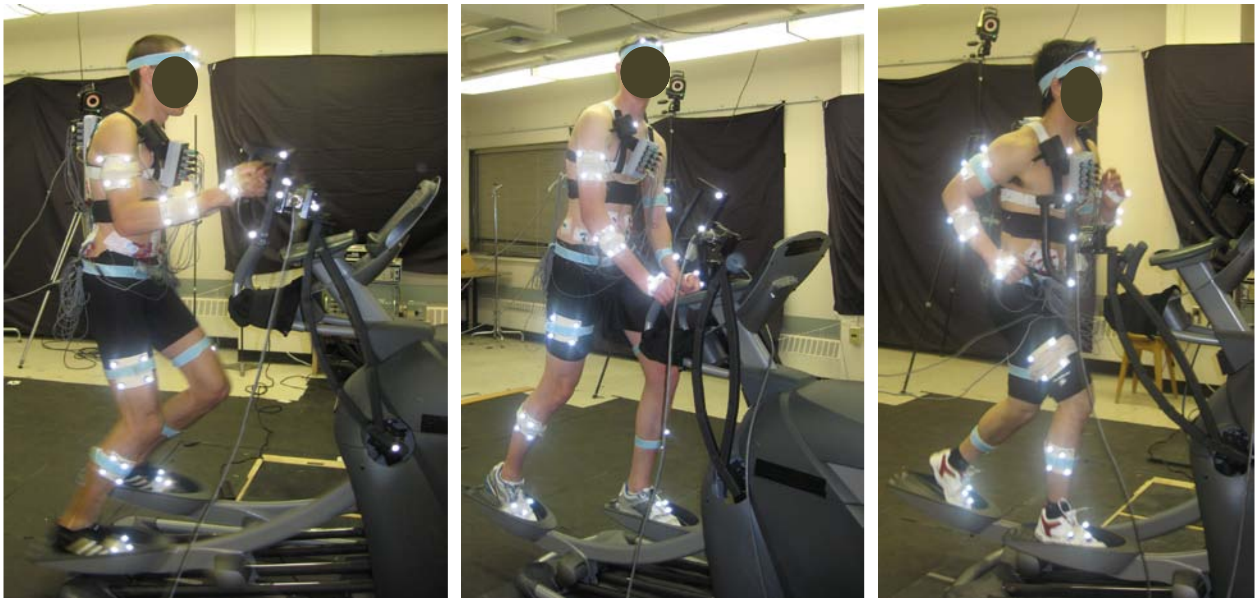


Figure 9-2: Photographs of participants on the elliptical trainer in the three positions tested. Left to right, they include: hands holding onto the handles, holding onto a central bar, and free-hand.

9.2.6. *Kinematics*

Motion data were processed using Visual 3D software (C-motion, Kingston, Ont, Canada). 3-dimensional lumbar and hip angles relative to the pelvis were calculated using a Visual 3D algorithm with a Cardan sequence of rotation (flexion/extension, side bending, followed by axial twist). Joint angles were filtered with a 6 Hz dual pass Butterworth filter. Signals were screened for abnormalities, processing errors, and marker movement. Maximum and minimum joint angles were taken from the entire capture time, unless the signal drifted over time due to body position changes (i.e. neck flexion, which tended to increase lumbar flexion), in which case the max/min were extracted from a complete cycle deemed representative of the normal scope of motion. To calculate average joint positions for lumbar and hip flexion angles, trials were clipped to ensure complete cycles of motion (similar to heel strike to heel strike in gait studies).

9.2.7. Kinetics

The data from 4 participants were used to calculate lumbar spine compression and shear forces. Prior to being used for kinetic calculations, force and EMG data were down-sampled to 60 Hz, so as to align with the marker position data. A top down rigid linked-segment model was constructed in Visual 3D. Standard inverse dynamics calculations were carried out to yield resulting forces and moments at the L4, L5 joint. Forces and moments from the two force cubes on the handles of the elliptical trainer were applied mid-hand. Outputs from these analyses, combined with data from the EMG were used to drive an anatomically detailed spine model representing 118 muscle fascicles as well as lumped parameter passive tissues, spanning T12 – L5, S1. This model has been described in detail previously (Cholewicki & McGill, 1996; Kavcic, Grenier, & McGill, 2004b; Moreside et al., 2006). Briefly, using the instantaneous spine position and EMG data, the model calculates individual muscle forces and stiffness, as well as the passive components (due to non-contractile tissues), to provide an internal moment. This is balanced with the external moment from a rigid linked segment model (from Visual 3d), using a least squared error method to calculate a gain factor. This gain is then applied to the internal forces. Total L4, L5 bone-on-bone forces are therefore the sum of the forces due to the external and gained internal moments.

9.2.8. Statistical analysis

All analyses utilized the SPSS (version 17) package with a significance level chosen at $p < 0.05$. For those kinematic and EMG data which had both right and left sided results, symmetry was assumed, and the right side was used for analysis. For lumbar angles and hip angles, a 2 X 2 X 3 repeated measures ANOVA was done with speed, stride length and hand position as the independent variables. Bonferroni adjustments were used to account for multiple comparisons, resulting in a significance level of $p < 0.05/n$ where n represents the number of comparisons within the selected group. Thus, for lumbar or hip angles: $n = 4, p < 0.0125$, and EMG for eight muscles: $n = 8, p < 0.006$. A repeated measures one-way ANOVA with simple contrasts and Bonferroni adjustment was used to compare lumbar motion in walking with that elicited in the 12 elliptical conditions. The same process was used to compare walking EMG levels with the 12 elliptical conditions, for each of the muscles being analyzed. Pearson correlations were used to compare lumbar angles on the elliptical with those elicited during walking, using the 4 elliptical trials which demonstrated the largest magnitude for each factor. Pearson correlations were also used to compare the 4 lumbar angles of interest with numerous anthropometric variables and speed.

9.3. Results

The elliptical trainer data from 3 participants was removed: two due to poor coordination and unsteadiness on the machine, one due to extensive marker position artifact.

9.3.1. *Elliptical vs. Walking*

Exercising on the elliptical trainer is not the same as walking. Participants tended to side bend more during walking, while twisting less (Figure 9-3). Although the average angle of forward lean was greater on the elliptical, the total amount of flexion/extension was only noticeably greater than walking in the fast speed and longer stride length condition (26”fast), where the average lean was 9.8°, compared to 7.8° in walking.

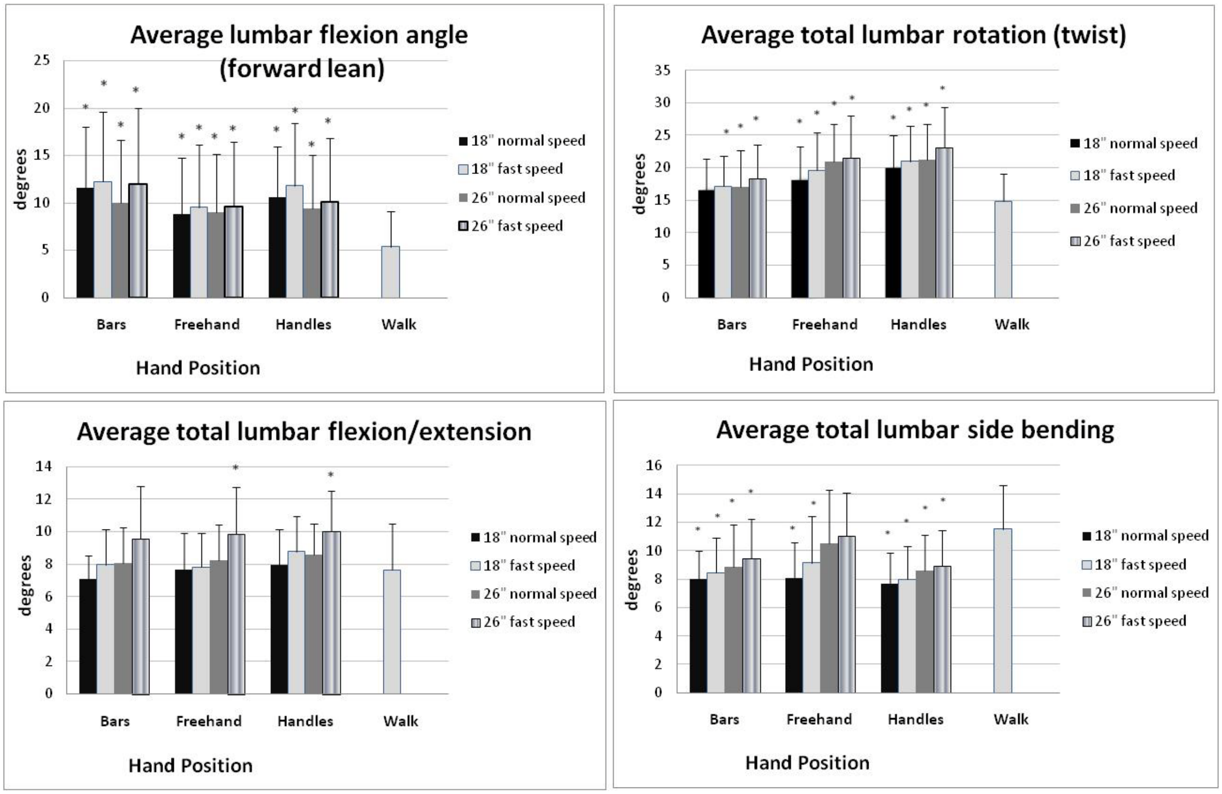


Figure 9-3: Lumbar motion on the elliptical trainer with varying speeds, stride lengths, and hand positions is compared to lumbar motion during walking. * = significantly different than walking, $p < 0.004$ (n = 37)

Peak trunk muscle activation was greater on the elliptical trainer for all of the muscles being analyzed, with the degree of difference varying according to speed, hand position, and stride length. The greatest difference was observed in the gluteal muscles, where average peak activations as high as 51% occurred in Glut Med in the fast 26" freehand condition, compared to 17% in walking (Figure 9-4).

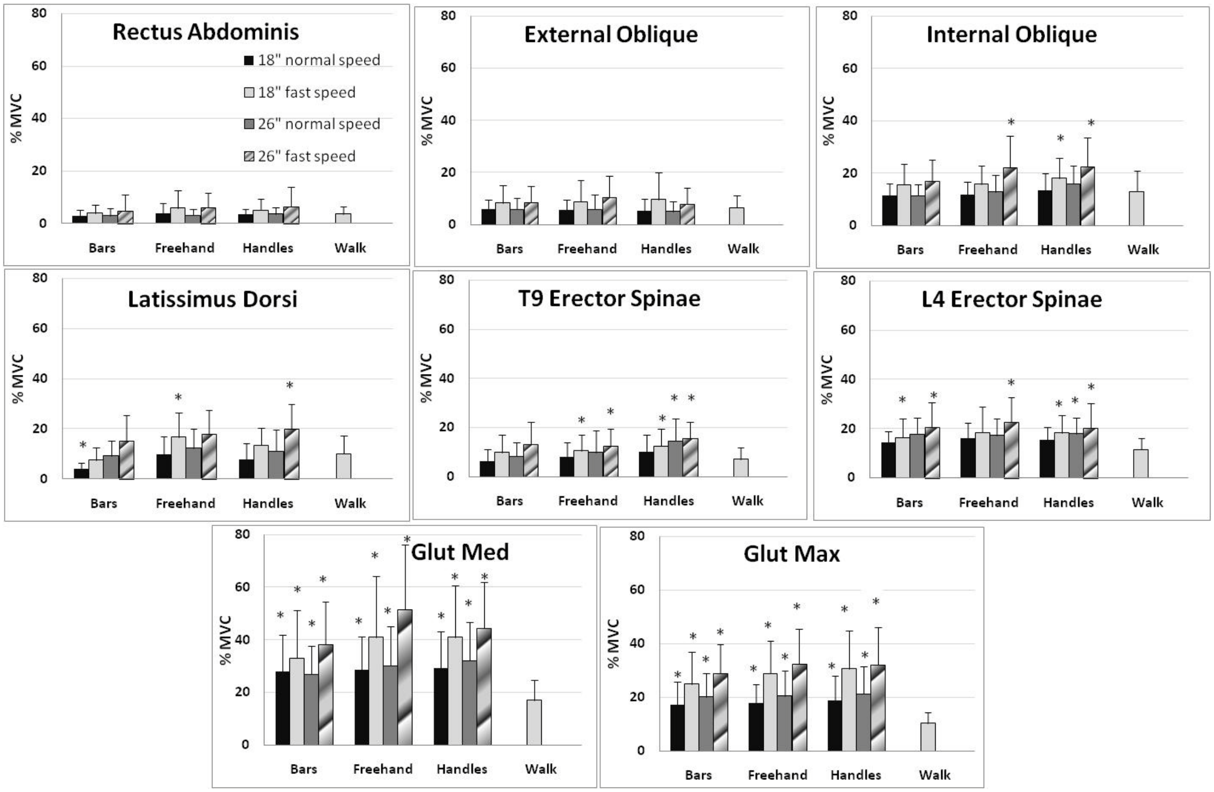


Figure 9-4: Trunk and gluteal peak muscle activation patterns comparing those occurring on the elliptical with those produced in walking. * = significantly different than walking, $p < 0.004$ (n = 20)

Furthermore, pearson correlation analyses demonstrated little relationship between lumbar motion occurring during walking compared to that taking place on the elliptical, other than in lumbar twist (Table 9-1). This motion showed a correlation of 0.826, or an r^2 value of 0.68, suggesting that the amount of rotation utilized by people in normal walking will be moderately predictive of how much they will rotate on the elliptical trainer. Correlations between walking and the elliptical for sagittal or frontal spine motion resulted in r^2 values of less than 0.19, thus poorly predictive.

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Table 9-1: Pearson correlations between mean lumbar motion occurring in walking compared to the elliptical trainer. W.xAvg = walking, average flexion; W.xTot = walking, total flexion/extension; W.yTot = walking, total side bending; W.zTot = walking, total lumbar twist; 18BNxAvg = average flexion with 18" bars, normal speed condition; 26HFxTot = total flexion/extension with 26" handles, fast speed; 26HFzTot = total lumbar twist with 26" handles fast; 26BNyTot = total side bending with 26" bars normal speed. Elliptical trials that demonstrated the greatest magnitude in each axis were chosen for comparisons. * = significant at the $p < 0.05$ level (n = 37)

	18BNxAvg	**W.xAvg**	**26HFxTot**	**W.xTot**	**26HFzTot**	**W.zTot**	**26BNyTot**	**W.yTot**
W.xAvg ***sig***	0.319 *0.055*							
26HFxTot ***sig***	0.006 *0.972*	0.210 *0.226*						
W.xTot ***sig***	-0.290 *0.082*	0.078 *0.648*	0.089 *0.611*					
26HFzTot ***sig***	-0.178 *0.305*	0.282 *0.096*	**0.411*** ***0.014***	0.243 *0.159*				
W.zTot ***sig***	-.190 *0.259*	-.007 *0.968*	**0.423*** ***0.011***	**0.410*** ***0.012***	**0.826*** ***0.000***			
26BNyTot ***sig***	-0.150 *0.383*	0.301 *0.074*	**0.428*** ***0.010***	0.129 *0.454*	0.130 *0.457*	0.139 *0.419*		
W.yTot ***sig***	-0.255 *0.127*	0.011 *0.947*	-0.244 *0.158*	**0.649*** ***0.000***	0.051 *0.770*	0.137 *0.419*	0.237 *0.164*	

Speed of elliptical cycle, stride length and hand position affect forward trunk lean and total spine motion about the three orthopaedic axes of flexion/extension, lateral bend and twist (Figure 9-3, Table 9-2). Specifically, an increase in speed resulted in greater trunk motion in all spine motions except lateral bend. Increased stride length resulted in a corresponding increase in total spine motion in all 3 axes, but did not significantly affect the average forward lean. The effect of hand position varied with the axis: the greatest lumbar flexion angle was elicited when holding onto the bars, the least with freehand. Both total flexion/extension and lumbar rotation increased from bars to freehand to handles, but only significantly so with rotation. Freehand elliptical resulted in the most lateral bending, with use of the handles being the least.

Table 9-2: Significance levels for the effect of speed, stride length and hand position on average lumbar spine motion. * indicates significance at the 0.05 level. (n = 37)

	Speed	Stride length	Hand position	Significant interactions
Average lumbar flexion angle	0.005*	0.070	0.001*	Stride/Hand: 0.000*
Total lumbar flexion/extension	0.000*	0.000*	0.343	--------
Total lumbar lateral bend	0.074	0.000*	0.000*	Stride/Hand: 0.005*
Total lumbar twist(rotation)	0.001*	0.000*	0.000*	--------

The effect of elliptical mechanical variables on sagittal hip mechanics is substantial. Namely, a longer stride length increases peak hip flexion and extension angles, resulting in greater flexion/extension excursion (Figure 9-5, Table 9-3). However, a change in speed

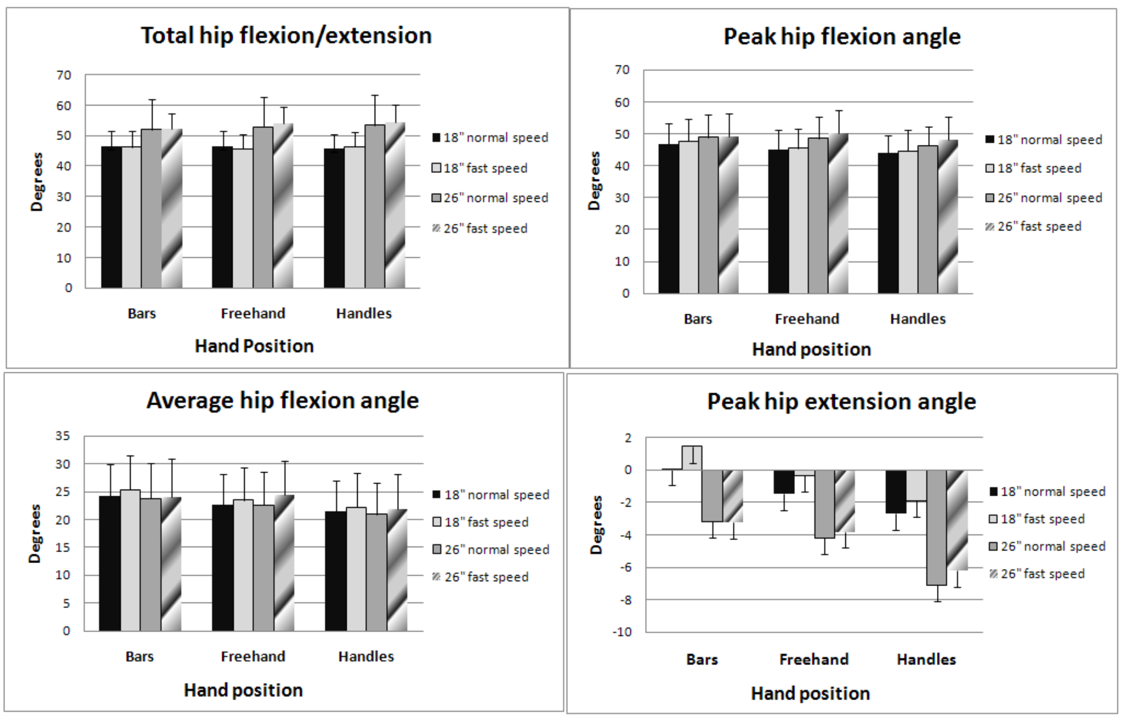

Figure 9-5: Average sagittal hip motion on the elliptical trainer, with varying speeds, stride lengths and hand positions. (n = 37)

forms an interesting situation. At the longer stride length, but normal speed, the largest hip extension angles were observed (average = -4.2°, collapsed across stride length and hand position). Adding more speed reduced hip extension to an average of -2.7° (Figure 9-5).

Table 9-3: Significance levels for the effect of speed, stride length and hand position on average hip sagittal motion. * indicates significance at the 0.05 level. (n = 37)

	Speed	**Stride length**	**Hand position**
Average hip flexion angle	0.005*	0.372	0.000*
Total hip flexion/extension	0.856	0.000*	0.073
Peak hip flexion	0.011*	0.000*	0.000*
Peak hip extension	0.000*	0.000*	0.000*

Anthropometric characteristics affected forward trunk lean and lumbar twist magnitudes. As seen in Table 9-4, taller participants demonstrated less lumbar twist, but a greater average flexion angle. For example, the tallest 5 participants demonstrated an average of 22.0(5)° of lumbar twist, compared to 28.1(6)° averaged over the shortest 5 (26HF condition). Similarly, these same 5 tall men averaged a forward lean angle of 18.4(8)° compared to 9.4(7)° in the shorter group (18BN condition). Measured passive hip extension also affected body mechanics on the elliptical: the greater the hip extension (which would be indicated by a negative number), the less lumbar rotation that occurred. There were no significant correlations between anthropometrics and total lumbar lateral bend or total flexion/extension.

Table 9-4: Pearson correlation results (r-values) and significance level inputting body anthropometrics and hip extension measurements. * = significance at the $p < 0.05$ level (n = 37)

	Arm Length	**Leg Length**	**Height**	**Avg. hip Ext.**
Total lumbar rotation (26HF) ***sig***	-0.355 *0.034*	-0.384* *0.021*	-0.460* *0.005*	0.374* *0.025*
Average lumbar flexion ***sig***	0.146 *0.194*	0.237 *0.158*	0.335* *0.042*	0.268 *0.109*

Muscle activation is different on the elliptical than when walking, and the stride length, speed and hand position also influence muscle activity. As shown in Figure 9-4, the gluteal muscles demonstrated the greatest activity, and the largest difference from those found in walking. Figure 9-6 demonstrates the high phasic activity found in those four muscles with the highest activation levels. All 16 muscles demonstrated average minimum %MVCs that were less than those demonstrated by L4ES (2.8%), confirming phasic activity across all muscles.

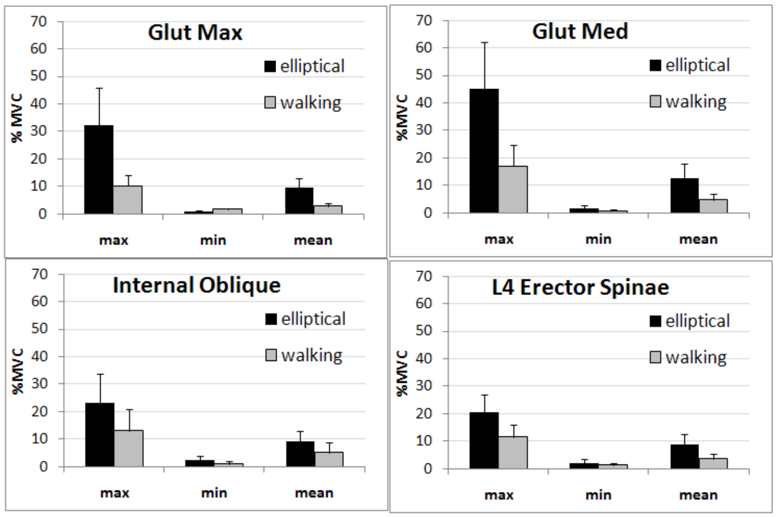

Figure 9-6: % MVC levels of the 4 muscles that demonstrated the highest activation levels on the elliptical trainer, compared to those found in walking. Maximum, minimum and mean %MVC are represented, and demonstrate the large phasic range elicited on the elliptical trainer. (n = 20)

Lumbar compression and shear forces were calculated on 4 participants using the elliptical trainer in an 18” stride, handles, normal speed condition (Table 9-6). For comparison sake, the forces were also normalized to body mass, as had been previously shown by Callaghan et al (1999) (Table 9-6). The resulting vertical compressive forces are similar to those found in walking at a fast cadence (20 steps/minute faster than self-selected). The anterior/posterior shear forces on the elliptical do not oscillate around zero, as is the case with walking, but instead result in a posterior shear of the thorax segment on the pelvis, due to the increased muscular activation of the back extensor muscles (Figure 9-4). However, the scope of the shear in both A/P and medial/lateral is larger on the elliptical than that seen in walking, likely a result of the exaggerated arm motion, lumbar rotation, and in some instances a larger scope of flexion/extension (Figure 9-3).

Table 9-5: Lumbar forces (N) calculated using an EMG driven model. Forces represent total bone-on forces: the sum of forces due to rigid linked segment modeling and those representing the force due to the muscle/soft tissue components. Negative ant/post shear forces indicate backward shear of the thorax relative to the pelvis.

	Compression			Ant/Post shear			Med/Lat shear		
subj	avg	max	min	avg	max	min	avg	max	min
11	1887	2897	997	-236	-435	-64	-8	167	-174
26	1556	2235	1016	-209	-465	-101	4	162	-124
34	1660	2138	1275	-192	-418	-38	66	145	-6
40	1402	1755	1049	-173	-286	-64	-16	63	-118
mean	**1626**	**2256**	**1084**	**-203**	**-401**	**-67**	**12**	**134**	**-106**
SD	204	475	129	27	79	26	37	48	71

Table 9-6: Average lumbar forces from **Error! Reference source not found.**, normalized to body mass (N/kg). Negative ant/post shear forces indicate backward shear of the thorax relative to the pelvis. The forces from the walking trials are those calculated by Callaghan et al (1999), using the same EMG driven model, but calculating joint reaction forces using “bottom up” approach, incorporating force plate data.

Condition		Compression			Ant/Post shear			Med/Lat shear		
		avg	max	min	avg	max	min	avg	max	min
elliptical: 18HN	mean	18.9	26.4	12.5	-2.4	-4.7	-0.8	0.1	1.6	-1.3
	SD	3.7	7.6	1.0	0.5	1.3	0.4	0.4	0.7	0.9
walking:	mean		21.8	12		1.2	-1.2		0.8	-0.7
normal speed	SD		4.6	2.2		0.3	0.4		0.3	0.2
walking:	mean		24.5	11.3		1.6	-1.6		1	-0.8
fast speed	SD		4.9	2		0.5	0.6		0.6	0.4

Discussion

There are many kinematic differences between walking and the elliptical trainer, with regard to the lumbar spine. Although the elliptical tends to constrain lumbar lateral bend, lumbar rotation and forward flexion were both dramatically increased on the elliptical (Figure 9-3). These motions are known to be causative in lumbar disc degeneration, annular delamination and facet encroachment. (Callaghan & McGill, 2001; Drake & Callaghan, 2009; Marshall & McGill, 2010). Given that the participants in this study averaged a cadence of

53(7) cycles per minute, or 69(9) at the faster speed, the total number of spine flexion/rotation events in a half hour session be 3180 - 4140, and considerably higher in those exercising at a faster than average velocity. The scope of axial rotation required on the elliptical when using the handles was 23°; almost identical to the 24° described in running by Schache et al (2003), but with an increased lumbar flexion angle, which differs from the generally extended position of the spine and anterior pelvic tilt in running (Franz, Paylo, Dicharry, Riley, & Kerrigan, 2009; Schache et al., 2003). Thus, the elliptical causes the spine to rotate through most of its available range of axial rotation but in an associated position of lumbar flexion.

Lumbar compression and shear forces are greater on the elliptical than those found in walking. Posterior shear of the trunk with respect to the pelvis, as demonstrated in Table 9-5 and Table 9-6, will tend to reduce the amount of contact between the lumbar facet joints, lessening force transmission through these joints. In turn, this may result in increased axial force being transmitted through the intervertebral discs, and increased available axial rotation. Consequently, people with discogenic low back pain may find that the elliptical trainer provokes their pain, due to the increased lumbar flexion angle as well as the lessening of facet joint contact due to the muscular-induced posterior shear of the thorax. Similarly, people with lumbar hypermobilities, who tend to rely on facet joint encroachment for force transmission and clinical stability, may find that exercising on the elliptical trainer results in increased lumbar rotation, past the point of comfort.

Hand position, stride length and velocity all had significant effects on lumbar spine kinematics (Figure 9-3). Those people with flexion intolerance would be encouraged to avoid holding onto the central bar for support, as it encourages a more flexed posture of the lumbar spine. However, it also demands the least lumbar twist, which may be advantageous for others. Increasing speed and stride length will, in general, produce the largest amount of spine rotation and flexion/extension, thus should be used with caution. Of interest, use of the elliptical in a 26” handles/fast condition resulted in average total lumbar rotation of 23.2°, yet voluntary active lumbar rotation in upright standing was only 23.9°. However, the lumbar flexion associated with this twist is greater on the elliptical, allowing opening of the facet joints and lessening the rotational compressive forces on these joints, albeit perhaps at the cost of increased annular stresses (Figure 9-7). Peak hip extension values are not high on the elliptical, especially when holding onto the bars. Normal walking resulted in an average of 13° of hip extension in our study, whereas the largest amount of average hip extension on the elliptical was 7° (Figure 9-5). Consequently, patients with limited hip extension may find the elliptical to be a satisfactory replacement for overground walking or treadmill use,

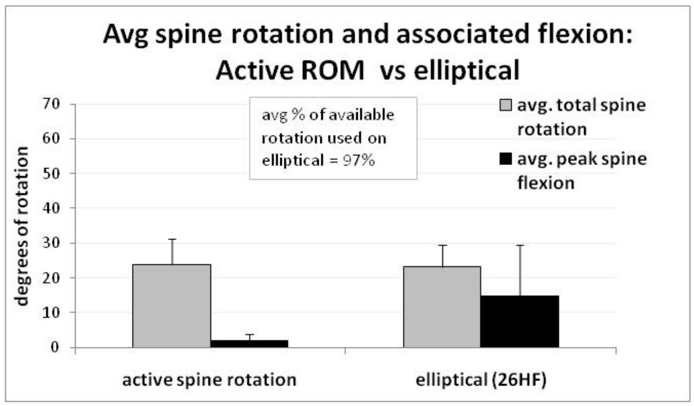

Figure 9-7: Average total lumbar rotation on the elliptical is nearly equal to that produced in upright standing, voluntary rotation. Associated lumbar flexion is greater on the elliptical. (n = 37)

which require hip extension values of 11° - 14°, regardless of speed (Franz et al., 2009; Riley et al., 2008; Schache et al., 2000). A person’s height will also affect lumbar motion on the elliptical. Despite the ability to vary stride length and hand position, there is no ability to raise or lower the handles or bars. Consequently, taller people may tend to flex more, yet rotate less, while the shorter people tend to do the opposite: adopt a more upright stance but rotate more around the vertical axis. One specific participant, who regularly participated in power lifting and weight training activities, demonstrated lumbar rotation measurements of approximately 25° to each side, with a resulting 51° of total lumbar rotation. Despite being only 172 cm in height, he had an average forward lean angle of 15°. However, this participant also demonstrated marked lack of hip mobility: 42° and 45° of total hip rotation (IR + ER) for the right (Rt) and left (Lt) leg, respectively, compared to a mean of 59.5° for the entire group. Consequently, he may have adopted this motor pattern of lumbar rotation to compensate for lack of hip mobility. One cannot ignore the effect of sport-specific training on spine kinematics, either. Although not addressed in this study, there did appear to be a trend for men who were highly trained in body contact sports, such as American football defensive players, to adopt a more flexed posture under activity, while those who specialized in running (distance or sprint) were likely to be more upright.

The highly phasic nature of the elliptical trainer promotes oxygenation and blood supply to the exercising muscles. Glut Med and IO are important stabilizers of the hip and spine, respectively (Kendall & McCreary, 1983; S. McGill, 2002; Sahrmann, 2002), and their activation levels on the elliptical indicate that it may be a superior choice compared to other exercise modalities. Cycling, although low impact, requires little gluteal activity: approximately 5% and 15% of MVC for GMax and GMed, respectively (Ericson, Nisell, Arborelius, & Ekholm, 1985; Ericson, 1988). These are relatively low when compared to the respective peak activation levels of 32% and 45% MVC demonstrated on the elliptical.

This study agrees with that of Burnfield (2010) and Lu (2007) who found increased spine and hip flexion on the elliptical, compared to walking. Both of these studies did not address different hand positions, speeds or stride lengths, and the focus of both was on lower extremity more so than the trunk. This may explain why Burnfield et al (2010) did not find a significant difference in GMed activity between walking and the elliptical, as our findings were that activity increased when the handles were not used, and stride length and speed were at their highest (Figure 9-3).

This study was limited to young, fit males, as it was part of a larger study that necessitated a younger group to reduce the likelihood of arthritic changes. Future studies should expand this information to include both sexes of varying ages. The results of this study represent the findings on only one type of elliptical trainer. There are many models available that may result in slightly different postures, and some with varying incline abilities. It was felt that variable stride length was an important feature to study, thus this specific model was chosen. Extrapolating these results to all models should be done with caution.

9.4. Summary

The elliptical trainer is different from walking. While not a substitute, it may be an exercise modality of choice for those desiring weight-bearing aerobic activity. Hip extension requirements are less than those in walking, yet there is significantly higher activity in the gluteal muscles, as well as the back extensors, LD and IO (depending on how the elliptical is used). Thus, it provides a strong phasic workout for the hip and trunk musculature, without stressing hip extension. However, exercising on the elliptical also results in a significant increase in the lumbar flexion angle and lumbar compression and rotation, while lessening the ability of the facet joints to transmit axial forces, compared to walking. Consequently, those people who are flexion and/or compression intolerant in the lumbar spine should be aware of these differences. Similarly, changing the stride length, velocity, and hand position on the elliptical will significantly affect the lumbar motion and muscle activity. Lumbar rotation is greatest when holding the handles, yet the average lumbar flexion angle tends to be highest when holding onto the central bar. Total lumbar motion in all 3 directions increases with stride length and velocity, but average lumbar flexion angle does not follow this pattern: when using the bars or handles, the spine is more flexed at a smaller stride length. Finally, there is a strong correlation between lumbar rotation in walking and that found on the elliptical. Thus, if a person tends to excessively rotate their lumbar spine in walking, this motion with be even more exaggerated on the elliptical trainer.

10. General Summary

This body of work has addressed a number of questions and hypotheses, adding to the literature regarding the relationship between hip mobility and the lumbar spine.

10.1. What constitutes limited hip mobility in a young adult male population?

77 young adult males participated, resulting in a normal distribution for hip extension, IR, ER and total hip rotation. This also produced percentile data which can be used to quantify limited and excessive hip mobility, providing measurements are obtained using the same participant positions. While it is generally accepted that arthritic hips lose mobility in a set pattern of restriction (Magee, 1987), this was not the case with this young population. Limitation of hip rotation correlated poorly with limitation of extension (r^2 value = 0.11). In turn, this had an effect on our participant recruitment in the other data collections: the frictionless jig study was originally to be one group of participants, with limited motion in both directions. Instead, it became two separate studies with two populations: those with extreme motion in rotation or extension, but seldom both. Similarly, it made it more difficult to find participants who demonstrated either excessive or limited mobility in both directions, as was required for the movement study comparing LHM and EHM groups (chapter 5).

10.2. Do young adult males with limited hip mobility move differently than those with normal or excessive ROM?

In common functional movements, such as walking, lunging, and twisting, it was difficult to demonstrate definitive differences between the two groups, as variability in movement patterns was high. The sagittal orientation of the pelvis was different between the groups, with the LHM group demonstrating more of an anterior tilt. Both the hip and lumbar spine angles are zero'd relative to the pelvis position in upright standing in most data collections modeling the hip and spine in-vivo, thus this difference in pelvis orientation may subsequently become hidden in ongoing trials. During functional movements, trends emerged demonstrating that the LHM group tended to flex, extend, and/or rotate their spine more than the EHM group, while latter would rely more on hip motion (Figure 5-9), but no statistical significance. The elliptical trainer was a useful medium to help constrain much of the variability seen in the other movement trials. It emerged that the LHM group adopted a posture of increased lumbar flexion, which would have an effect of lessening the amount of absolute lumbar extension occurring with each cycle. While it was anticipated that they might demonstrate increased "hinging" into extension than the EHM group, this was not necessarily the case in the active trials. Passively, however, this was demonstrated on the frictionless jig (chapter 8). When an extension moment was applied to the distal leg, the LHE group showed significantly less resistance to movement in the spine than did the EHM group. Thus, it appears that in this group of young men, the passive tissues are less able to resist a lumbar extension moment, but the active contractile tissues compensate during movement: supporting the spine and positioning it to limit absolute extension. Potentially, if motor control was less than optimal, such as in an unhealthy or low back pain population, the contractile tissues may not be able to protect the spine, and this decreased passive resistance to extension could result in excessive lumbar motion and injury.

In that much of this thesis compared a group with LHM to one with EHM, the question arises as to which would be preferable, given the results of this research. This may depend on each individual's activity requirements; while a runner may prefer hips with limited mobility, to maximize elastic energy conserved in the tendinous and ligamentous structures, a gymnast or hurdler would likely choose excessive mobility to enable extreme positions of the hip joint. While it was anticipated that the group with LHM would demonstrate increased lumbar motion, there was little definitive evidence of this in our young, healthy population. The group with EHM utilized more gluteal muscle activation during elliptical use, which is in keeping with previous research indicating increased metabolic efficiency in a group with limited lower limb flexibility (Gleim et al, 1990). As demonstrated in Chapter 6, it is possible to enhance hip mobility with a 6 week exercise/stretching protocol. Conversely, there appears to be no scientific literature describing a protocol to reduce motion in a joint with excessive mobility. Anecdotally, it is much more difficult to stabilize a hyper-mobile joint with an exercise regime than to increase ROM in a joint with limited mobility in a non-surgical rehabilitation setting. Thus, it would appear that, although ROM which is closer to the 50^{th} percentile might be ideal for normal activity, presenting with hip mobility which is less than average would be preferable to having excessively mobile hips for most people.

10.3. In a group of young adult males with limited hip mobility, is it possible to enhance their hip range of motion and/or core strength and endurance with a 6 week exercise protocol? Do such gains subsequently result in changes to movements patterns in the hip and back?

Six weeks of stretching resulted in large increases in measured ROM, and a similar improvement in muscle endurance. Equally so, motor control exercises were progressed in difficulty, based on a criterion of being able to perform the activity with minimal concurrent spine motion. However, despite all of these quantifiable improvements, participants showed very little definitive change in how they moved in the lumbar spine and hip during the functional movements chosen for analysis. The group that received only hip stretching demonstrated an interesting tendency to actually increase their lumbar motion and decrease their hip motion in most of the conditions. This is a concept worthy of further study: although sports and rehabilitation professionals often focus on stretching tissues with limited extensibility, improvements are not necessarily transferred to function. In fact, depending on what motor control education is associated with the exercise routine, the focus of the movement may be misconstrued, resulting in even more aberrant movement patterns than those initially observed. Human movement is a careful balance of available ROM and tissue tensions, strength, motor control, joint and whole body stability, as well as metabolic efficiency. We cannot assume that changing any one of these factors in isolation will result in an overall change to movement. As suggested in chapter 6, a concurrent increase in trunk muscle strength and/or endurance resulted a greater tendency of improved functional hip ROM with less spine motion, than did hip stretching alone (Figure 6-6 to Figure 6-11). Changing motion needs to be approached, not only in terms of anatomical structures, but the appropriate recruitment and motor control of these tissues (Caillou et al., 2002; Delignières et al., 1998). Old patterns of movement must be overcome in order to learn new patterns that exploit the changes in tissue tension gained with an exercise program (Nourrit et al., 2003; Vereijken et al., 1997).

10.4. How do the passive stiffness qualities differ in these two groups?

In a frictionless environment, young men with limited hip mobility tend to adopt a hip rotational posture which differs from their colleagues with excessive mobility. In upright standing, they stand with the hips more externally rotated, potentially reducing tension on the posterior hip structures, while transferring this tension anteriorly. This shift might reduce the necessity of muscular activity: anterior hip stability is generally considered to be controlled by the strong Y-ligament, whereas the posterior hip relies more on muscular control. Subsequent to these posture changes, the anterior hip structures, when pulled into ER, were dramatically stiffer than that of the EHR group (Figure 7-11), likely due to the ligamentous resistance. An IR moment did not demonstrate such a large differential between groups. The fact that the LHM group was mildly stiffer throughout range suggests that the difference may be due to musculo-tendinous structures, as opposed to ligamentous, the latter being more likely to come into effect at the end of range.

In side lying, a similar trend was demonstrated with the anterior hip structures: greater stiffness in the LHE group than the EHE, although there was no difference in their initial starting position when the knee was extended. Flexing the knee in both groups lessened the stiffness, as resistance to an extension moment was transferred to the longer two-joint structures (rectus femoris, gracilis) as opposed to the anterior hip joint ligaments, psoas, and iliacus.

Of functional importance, the LHE group tended to extend their spine more than the EHE group for a given extension moment applied at the distal leg, indicating that they have less passive resistance to extension in the spine. Consequently, motor control to overcome this tendency becomes imperative to protect the spine from repetitive extremes of extension.

10.5. Final remarks: Exercise prescription

This new understanding of how limited hip mobility affects the lumbar spine underscores the importance of evaluating hip ROM as part of a clinical musculoskeletal assessment involving the lumbar spine. Not only do those young adults with LHM stand with more of an anterior tilt to the pelvis, but they demonstrate less passive resistance to extension in the lumbar spine when an extension moment is applied. Attempts at improving their range of hip extension and/or rotation, therefore, should be undertaken in conjunction with an appropriate spine stabilization protocol, preferably one that incorporates functional movement patterns, so that these preferred patterns become efficient and automatic.

Exercising on the elliptical trainer requires less hip extension than walking, yet results in a superior muscular challenge for the gluteal muscles, the upper and lower ES, LD and IO. However, elliptical use also results in increased lumbar rotation and flexion, as well as greater lumbar bone on bone compression and posterior shear. This increased posterior shear of the thorax on the pelvis will tend to lessen facet joint contact, thus increasing the percentage of axial compression being transmitted by the intervertebral disc. Thus, patients whose lumbar spine may be intolerant of an increase in these motions and compressive/shear forces are advised to use caution when using the elliptical. Specifically, slower speeds elicit less of the rotation and flexion, as well as lower muscle activation levels, which would result in less lumbar compression and shear. In addition, those people with limited hip mobility demonstrated an even greater lumbar flexion angle and lumbar compression on the elliptical trainer when compared to those with greater hip mobility. Thus, patients who present with limited hip mobility, who are also intolerant of increased spine load, should be aware of these outcomes. The elliptical trainer may not be the exercise modality of choice for them.

References

Adams, M. A., Dolan, P., & Hutton, W. C. (1988). The lumbar spine in backward bending. *Spine, 13*(6-12), 1019-1026.

Adams, M. A., & Hutton, W. C. (1980). The effect of posture on the role of the apophyseal joints in resisting intervertebral compressive forces. *Journal of Bone and Joint Surgery, 62B*, 358-362.

Adams, M. A., & Hutton, W. C. (1981). The relevance of torsion to the mechanical derangement of the lumbar spine. *Spine, 6*(3), 241-248.

Adams, M. A., Hutton, W. C., & Stott, J. R. R. (1980). The resistance to flexion of the lumbar intervertebral joint. *Spine, 5*(3), 245-253.

Adams, M. A., May, S., Freeman, B. J., Morrison, H. P., & Dolan, P. (2000). Effects of backward bending on lumbar intervertebral discs. relevance to physical therapy treatments for low back pain. *Spine, 25*(4), 431-7; discussion 438.

Andersson, E. A., Oddsson, L. I. E., Grundstrom, H., Nilsson, J., & Thorstensson, A. (1996). EMG activities of the quadratus lumborum and erector spinae muscles during flexion-relaxation and other motor tasks. *Clinical Biomechanics, 11*(7), 392-400.

Andersson, E. A., Oddsson, L. I. E., Grundstrom, H., & Thorstensson, A. (1995). *The role of the psoas and iliacus muscles for stability and movement of the lumbar spine, pelvis and hip*

Aspden, R. M., Rudman, K. E., & Meakin, J. R. (2006). A mechanism for balancing the human body on the hips. *Journal of Biomechanics, 39*(9), 1757-1759.

Aultman, C. D., Scannell, J., & McGill, S. M. (2005). The direction of progressive herniation in porcine spine motion segments is influenced by the orientation of the bending axis. *Clinical Biomechanics (Bristol, Avon), 20*(2), 126-129.

Bach, D. K., Green, D. S., Jensen, G. M., & Savinar, E. (1985). A comparison of muscular tightness in runners and nonrunners and the relation of muscular tightness to low back pain in runners. *The Journal of Orthopaedic and Sports Physical Therapy, 6*, 315-323.

Bandy, W. D., & Irion, J. M. (1994). The effect of time on static stretch on the flexibility of the hamstring muscles. *Physical Therapy, 74*(9), 845-852.

Bandy, W. D., Irion, J. M., & Baltz, M. (1994). The effect of static stretch on the flexibility of the hamstring muscles as measured by knee extension range of motion. *Journal of Orthopaedic and Sports Physical Therapy, 19,* 72.

Bandy, W. D., Irion, J. M., & Briggler, M. (1997). The effect of time and frequency of static stretching on flexibility of the hamstring muscle. *Physical Therapy, 77*(10), 1090-1096.

Barbee Ellison, J. B., Rose, S. J., & Sahrmann, S. (1990). Patterns of hip rotation range of motion: A comparison between healthy subjects and patients with low back pain. *Physical Therapy, 70,* 537-541.

Beach, T. A. C., Parkinson, R. J., Stothart, J. P., & Callaghan, J. P. (2007). Effects of prolonged sitting on the passive flexion stiffness of the invivo lumbar spine. *The Spine Journal, 5*(2), 145-154.

Benedetti, M. G., Catani, F., Leardini, A., Pignotti, E., & Giannini, S. (1998). Data management in gait analysis for clinical applications. *Clinical Biomechanics*, *13*(3), 204-215.

Ben-Galim, P., Ben-Galim, T., Rand, N., Haim, A., Hip, J., Dekel, S., et al. (2007). Hip-spine syndrome: The effect of total hip replacement surgery on low back pain in severe osteoarthritis of the hip. *Spine, 32*(19), 2099-2102.

Benoit, D. L., Ramsey, D. K., Lamontagne, M., Xu, L., Wretenberg, P., & Renström, P. (2006). Effect of skin movement artifact on kne kinematics during gait and cutting motions measured in vivo. *Gait and Posture, 24*, 152-164.

Bigliani, L. U., Pollock, R. G., Soslowsky, L. J., Flatow, E. L., Pawluk, R. J., & Mow, V. C. (1992). Tensile properties of the inferior glenohumeral ligament. *Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society*, *10*(2), 187-197.

Björklund, M., Hamberg, J., & Crenshaw, A. G. (2001). Sensory adaptation after a 2-week stretching regimen of the rectus femoris muscle. *Arch Phys Med Rehabil, 82*, 1245-1250.

Bogduk, N., & Twomey, L. T. (1991). *Clinical anatomy of the lumbar spine*. Melbourne: Churchill Livingstone.

Bogduk, N. L. T. (1997). *Clinical anatomy of the lumbar spine and sacrum*. New York: Churchill Livingstone.

Booher, J. M., & Thibodeau, G. A. (2000). *Athletic injury assessment*. Boston: McGraw-Hill.

Boone, D. C., Azen, S. P., Lin, C., Spence, C., Baron, C., & Lee, L. (1978). Reliability of goniometric measurements. *Physical Therapy, 58*(11), 1355-1360.

Bovens, A. M. P. M., van Baak, M. A., Vrencken, J. G. P. M., Wijnen, J. A. G., & Verstappen F.T.J. (1990). Variability and reliability of joint measurements. *The American Journal of Sports Medicine, 18*(1), 58-63.

Brown, S. H. M., & McGill, S. M. (2008). How the inherent stiffness of the in-vivo human trunk varies with changing magnitudes of muscular activation. *Clinical Biomechanics, 23*, 15-22.

Bruijn, S., Lamoth, C., Kingma, I., Meijer, O., & van Dieen, J. H. (2006). Coordination between pelvis, thorax and leg movements in gait. *Gait and Posture, 24S*, S10-S11.

Burnfield, J. M., Josephson, K. R., Powers, C. M., & Rubenstein, L. Z. (2000). The influence of lower extremity joint torque on gait characteristics in elderly men. *Arch Phys Med Rehabil, 81*, 1153-1157.

Burnfield, J. M., Shu, Y., Buster, T., & Taylor, A. (2010). Similarity of joint kinematics and muscle demands between elliptical training and walking: Implications for practice. *Physical Therapy, 90*(2), 1-17.

Caillou, D., Nourrit, D., Deschamps, T., Lauriot, B., & Delignieres, D. (2002). Overcoming spontaneous patterns of coordination during the acquisition of a complex balancing task. *Canadian Journal of Experimental Psychology, 56*(4), 283-293.

Callaghan, J. P., & McGill, S. M. (1995). Muscle activity and low back loads under external shear and compressive loading. *Spine, 20*(9), 992-998.

Callaghan, J. P., & McGill, S. M. (2001). Intervertebral disc herniation: Studies on a porcine model exposed to highly repetitive flexion/extension motion with compressive force. *Clinical Biomechanics (Bristol, Avon), 16*(1), 28-37.

Callaghan, J. P., Patla, A. E., & McGill, S. (1999). Low back three-dimensional joint forces, kinematics and kinetics during walking. *Clinical Biomechanics, 14,* 203-216.

Caplan, N., Rogers, R. R., Parr, M. K., & Hayes, P. R. (2009). The effect of proprioceptive neuromuscular facilitation and static stretch training on running mechanics. *Journal of Strength and Conditioning Research, 23*(4), 1175-1180.

Cappozzo, A. (1983). The forces and couples in the human trunk during level walking. *Journal of Biomechanics, 16*(4), 265-277.

Cappozzo, A. (1984). Compressive loads in the lumbar vertebral column during normal level walking. *Journal of Orthopaedic Research, 1*, 292-301.

Cappozzo, A., Catani, F., Leardini, A., Benedetti, M. G., & Della Croce, U. (1996). Position and orientatin in space of bones during movement: Experimental artefacts. *Clinical Biomechanics, 11*(2), 90-100.

Chandrashekar, N., Mansouri, H., Slauterbeck, J., & Hashemi, J. (2006). Sex-based differences in the tensile properties of the human anterior cruciate ligament. *Journal of Biomechanics, 39*, 2943-2950.

Chandrashekar, N., Slauterbeck, J., & Hashemi, J. (2005). Sex-based differences in the anthropometric characteristics of the anterior cruciate ligament and its relation to intercondylar notch geometry: A cadaveric study. *The American Journal of Sports Medicine, 33*(10), 1492-1498.

Chesworth, B. M., Padfield, B. J., Helewa, A., & Stitt, L. W. (1994). A comparison of hip mobility in patients with low back pain and matched healthy subjects. *Physiotherapy Canada, 46*(4), 267-274.

Cholewicki, J., & McGill, S. (1996). Mechanical stability of the *in-vivo* lumbar spine: Implications for injury and chronic low back pain. *Clinical Biomechanics, 11*(1), 1-15.

Cholewicki, J., McGill, S., & Norman, R. W. (1995). Comparison of muscle forces and joint load from an optimization and EMG assisted lumbar spine model: Towards development of a hybrid approach. *Journal of Biomechanics, 28*(3), 321-331.

Cibulka, M. T. (1999). Low back pain and its relation to the hip and foot. *Journal of Orthopaedic and Sports Physical Therapy, 29*(10), 595-601.

Cibulka, M. T., Sinacore, D. R., Cromer, G. S., & Delitto, A. (1998). Unilateral hip rotation range of motion asymmetry in patients with sacroiliac joint regional pain. *Spine, 23*(9), 1009-1015.

Clayson, S. J., Newman I.M, Debevec, D. F., Anger R.W., Skowlund, H. V., & Kottke, F. J. (1962). Evaluation of mobility of hip and lumbar vertebrae of normal young women. *Archives of Physical Medicine and Rehabilitation, 43*, 1-8.

Crawford, M. J., Dy, C. J., Alexander, J. W., Thompson, M., Schroder, S. J., Vega, C. E., et al. (2007). The biomechanics of the hip labrum and the stability of the hip. *Clinical Orthopaedics and Related Research, 465*, 16-22.

Crosbie, J., Vachalathiti, R., & Smith, R. (1997a). Age, gender and speed effects on spinal kinematics during walking. *Gait and Posture, 5*, 13-20.

Crosbie, J., Vachalathiti, R., & Smith, R. (1997b). Patterns of spinal motion during walking. *Gait and Posture, 5*, 6-12.

Cyriax, J. (1975). *Textbook of orthopaedic medicine*. Baltimore: The Williams & Wilkins company.

Davis, R. B., Ounpuu, S., Tyburski, D., & Gage, J. R. (1991). A gait analysis data collection and reduction technique. *Human Movement Science, 10*, 575-587.

Delignières, D., Nourrit, D., Sioud, R., Leroyer, P., Zattara, M., & Micaleff, J. (1998). Preferred coordination modes in the first steps of the learning of a complex gymnastics task. *Human Movement Science, 17*, 221-241.

Dickey, J. P., McNorton, S., & Potvin, J. R. (2003). Repeated spinal flexion modulates the flexion-relaxation phenomenon. *Clinical Biomechanics, 18*, 783-789.

Dolan, P., & Adams, M. A. (1993). Influence of lumbar and hip mobility on the bending stresses acting on the lumbar spine. *Clinical Biomechanics, 8*, 185-192.

Drake, J. D., Aultman, C. D., McGill, S. M., & Callaghan, J. P. (2005). The influence of static axial torque in combined loading on intervertebral joint failure mechanics using a porcine model. *Clinical Biomechanics (Bristol, Avon), 20*(10), 1038-1045.

Drake, J. D., & Callaghan, J. P. (2009). Intervertebral neural foramina deformation due to two types of repetitive combined loading. *Clinical Biomechanics (Bristol, Avon), 24*(1), 1-6.

Dunk, N. M., & Callaghan, J. P. (2005). Gender-based differences in postural responses to seated postures. *Clinical Biomechanics, 20*, 1101-1110.

Dunk, N. M., Kedgley, A. E., Jenkyn, T. R., & Callaghan, J. P. (2009). Evidence of a pelvis-driven flexion pattern: Are the joints of the lower lumbar spine fully flexed in seated postures? *Clinical Biomechanics (Bristol, Avon), 24*(2), 164-168.

Dunlop, R. B., Adams, M. A., & Hutton, W. C. (1984). Disc space narrowing and the lumbar facet joints. *The Journal of Bone and Joint Surgery, 66B*(5), 706-710.

Dy, C. J., Thompson, M. T., Crawford, M. J., Alexander, J. W., McCarthy, J. C., & Noble, P. C. (2008). Tensile strain in the anterior part of the acetabular labrum during provocative maneuvering of the normal hip. *The Journal of Bone and Joint Surgery.American Volume, 90*(7), 1464-1472.

Eke-Okoro, S. T., Gregoric, M., & Larsson, L. E. (1997). Alterations in gait resulting from deliberate changes of arm-swing amplitude and phase. *Clinical Biomechanics, 12*(7), 516-521.

Ekstrand, J., Wiktorsson, M., Öberg, B., & Gillquist, J. (1982). Lower extremity goniometric measurements: A study to determine their reliability. *Arch Phys Med Rehabil, 63,* 171-175.

Elftman, H. (1939). The function of the arms in walking. *Human Biology, 11*, 529-535.

Ellenbecker, T. S., Ellenbecker G.A., Roetert, E. P., Silva, R. T., Keuter, G., & Sperling F. (2007). Descriptive profile of hip rotation range of motion in elite tennis players and professional baseball pitchers. *American Journal of Sports Medicine,* , in press.

Ericson, M. O. (1988). Muscular function during ergometer cycling. *Scandinavian Journal of Rehabilitation Medicine, 20*(1), 35-41.

Ericson, M. O., Nisell, R., Arborelius, U. P., & Ekholm, J. (1985). Muscular activity during ergometer cycling. *Scandinavian Journal of Rehabilitation Medicine, 17*(2), 53-61.

Esola, M. A., McClure, P. W., Fitzgerald, G. K., & Siegler, S. (1996). Analysis of lumbar spine and hip motion during forward bending in subjects with and without a history of low back pain. *Spine, 21*(1), 71-78.

Fairbank, J. C. T., Pynsent, P. B., Van Poortvliet, J. A., & Phillips, H. (1984). Influence of anthropometric factors and joint laxity in the incidence of adolescent back pain. *Spine, 9*, 461-464.

Farley, C. T., & Morgenroth, D. C. (1999). Leg stiffness primarily depends on ankle stiffness durfing human hopping. *Journal of Biomechanics, 32*, 267-273.

Feland, J. B., Myrer, J. W., Schulthies, S. S., Fellingham, G. W., & Measom, G. W. (2001). The effect of duration of stretching of the hamstring muscle group for increasing range of motion in people aged 65 years or older. *Physical Therapy, 81*, 1100-1117.

Fenwick, C. M., Brown, S. H., & McGill, S. M. (2009). Comparison of different rowing exercises: Trunk muscle activation and lumbar spine motion, load, and stiffness. *Journal of Strength and Conditioning Research / National Strength & Conditioning Association, 23*(5), 1408-1417.

Ferber, R., McClay Davis, I., & Williams, D. S. I. (2003). Gender differences in lower extremity mechanics during running. *Clinical Biomechanics, 18*, 350-357.

Folpp, H., Deall, S., Harvey, L. A., & Gwinn, T. (2006). Can apparent increases in muscle extensibility with regular stretch be explained by changes in tolerance to stretch? *Australian Journal of Physiotherapy, 52*, 45-50.

Franz, J. R., Paylo, K. W., Dicharry, J., Riley, P. O., & Kerrigan, D. C. (2009). Changes in the coordination of hip and pelvis kinematics with mode of locomotion. *Gait and Posture, 29*, 494-498.

Fuss, F. K., & Bacher, A. (1991). New aspects of the morphology and function of the human hip joint ligaments. *The American Journal of Anatomy, 192*(1), 1-13.

Gabbe, B. J., Bennell, K. L., Wajswelner, H., & Finch, C. F. (2004). Reliability of common lower extremity musculoskeletal screening tests. *Physical Therapy in Sport, 5*, 90-97.

Gajdosik, R. L. (1991). Effects of static stretching on the maximal length and resistance to passive stretch of short hamstring muscles. *Journal of Orthopaedic and Sports Physical Therapy, 14*(6), 250-255.

Garrett, W. E. J. (1990). Muscle strain injuries: Clinical and basic aspects. *Medicine and Science in Sports and Exercise, 22*(4), 436-443.

Gleim, G. W., Stachenfeld, N. S., & Nicholas, J. A. (1990). The influence of flexibility on the economy of walking and jogging. *Journal of Orthopaedic Research, 8*, 814-823.

Gombatto, S. P., Collins, D. R., Sahrmann, S., Engsberg, J. R., & Van Dillen, L. R. (2006). Gender differences in pattern of hip and lumbopelvic rotation in people with low back pain. *Clinical Biomechanics, 21*, 263-271.

Gossman, M. R., Sahrmann, S., & Rose, S. J. (1982). Review of length-associated changes in muscle. *Physical Therapy, 62*(12), 1799-1808.

Gracovetsky, S. (1997). Linking the spinal engine with the legs: A theory of human gait. In A. Vleeming, V. Mooney, T. Dorman, C. Snijders & R. Stoeckart (Eds.), *Movement, stability and low back pain* (pp. 243-251). Edinburgh: Churchill Livingstone.

Granata, K. P., & Marras, W. S. (1993). An EMG assisted model of loads on the lumbar spine during asymmetric trunk extensions. *Journal of Biomechanics, 26*(12), 1429-1438.

Gray, H. (1974). In Pick T. P., Howden R. (Eds.), *Gray's anatomy*. Philadelphia, PA: Running Press.

Grimshaw, P. N., & Burden, A. M. (2000). Case report: Reduction of low back pain in a professional golfer. *Medicine and Science in Sports and Exercise, 32*(10), 1667-1673.

Grood, E. S., & Suntay, W. J. (1984). A joint coordinate system for the clinical description of three dimensional motions: Application to the knee. *Journal of Biomechanical Engineering, 105*, 136-144.

Growney, E., Meglan, D., Johnson, M., Cahalan, T., & An, K. (1997). Repeated measures of adult normal walking using a video tracking system. *Gait and Posture, 6*, 147-162.

Gupta, A. (2001). Analyses of myo-electric silence of erectors spinae. *Journal of Biomechanics, 34,* 491-496.

Hamill, J., Moses, M., & Seay, J. (2009). Lower extremity joint stiffness in runners with low back pain. *Research in Sports Medicine (Print), 17*(4), 260-273.

Hardcastle, P., Annear, P., Foster, D. H., Chakera, T. M., McCormick, C., Khangure, M., et al. (1992). Spinal abnormalities in young fast bowlers. *Journal of Bone and Joint Surgery, 74(B)*(3), 421-425.

Harvey, D. (1998). Assessment of the flexibility of elite athletes using the modified thomas test. *British Journal of Sports Medicine, 32*, 68-70.

Harvey, L., Herbert, R., & Crosbie, J. (2002). Does stretching induce lasting increases in joint ROM? A systematic review. *Physiotherapy Research International, 7*(1), 1-13.

Hewitt, J., Guilak, F., Glisson, R., & Vail, T. P. (2001). Regional material properties of the human hip joint capsule ligaments. *Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society, 19*(3), 359-364.

Hewitt, J. D., Glisson, R. R., Guilak, F., & Vail, T. P. (2002). The mechanical properties of the human hip capsule ligaments. *The Journal of Arthroplasty, 17*(1), 82-89.

Hidaka, E., Aoki, M., Muraki, T., Izumi, T., Fujii, M., & Miyamoto, S. (2009). Evaluation of stretching position by measurement of strain on the ilio-femoral ligaments: An in vitro simulation using trans-lumbar cadaver specimens. *Manual Therapy, 14*(4), 427-432.

Hobara, H., Kimura, K., Omuro, K., Gomi, K., Muraoka, T., Sakamoto, M., et al. (2010). Differences in lower extremity stiffness between endurance-trained athletes and untrained subjects. *Journal of Science and Medicine in Sport / Sports Medicine Australia, 13*(1), 106-111.

Hoppenfeld, S. (1976). *Physical examination of the spine and extremities*. New York, NY: Appleton-Century-Crofts.

Huijing, P. A. (2007). Muscle as a collagen fiber reinforced composite: A review of force transmission in muscle and whole limb. *Journal of Biomechanics, 32,* 329-345.

Huijing, P. A. (2009). Epimuscular myofascial force transmission: A historical review and implications for new research. international society of biomechanics muybridge award lecture, taipei, 2007. *Journal of Biomechanics, 42*(1), 9-21.

Hurd, W. J., Chmielewski, T. L., Axe, M. J., Davis, I., & Snyder-Mackler, L. (2004). Differences in normal and perturbed walking kinematics between male and female athletes. *Clinical Biomechanics, 19*, 465-472.

James, B., & Parker, A. W. (1989). Active and passive mobility of lower limb joints in elderly men and women. *American Journal of Physical Medicine and Rehabilitation, 67-68,* 162-167.

Johnston, R. C., & Smidt, G. L. (1969). Measurement of hip-joint motion during walking. *Journal of Bone and Joint Surgery (American Ed.), 51*(6), 1082-1094.

Juker, D., McGill, S., Kropf, P., & Steffen, T. (1998). Quantitative intramuscular myoelectric activity of lumbar portions of psoas and the abdominal wall during a wide variety of tasks. *Med Sci Sports Exercise*, *30*(2), 301-310.

Kanayama, M., Abumi, K., Kaneda, K., Tadano, S., & Ukai, T. (1996). Phase lag of the intersegmental motion in flexion-extension of the lumbar and lumbo-sacral spine: An in vivo study. *Spine, 21*, 1416-1422.

Karamanidis, K., & Arampatzis, A. (2007). Aging and running experience affects the gearing in the musculoskeletal system of the lower extremities while walking. *Gait and Posture, 25*, 590-596.

Kavcic, N., Grenier, S. G., & McGill, S. M. (2004a). Quantifying tissue loads and spine stability while performing commonly prescribed low back stabilization exercises. *Spine, 29*(20), 2319-2329.

Kavcic, N., Grenier, S. G., & McGill, S. (2004b). Determining the stabilizing role of individual torso muscles during rehabilitation exercises. *Spine, 29*(11), 1254-1265.

Kendall, F. P., & McCreary, E. K. (1983). *Muscles, testing and function, 3rd ed*. Baltimore: Williams & Wilkens.

Kerrigan, D. C., Lee, L. W., Collins, J. J., Riley, P. O., & Lipsitz, L. A. (2001). Reduced hip extension during walking: Healthy elderly and fallers versus young adults. *Archives of Physical Medicine and Rehabilitation, 82,* 26-30.

Kerrigan, D. C., Todd, M. K., & Della Croce, U. (1998). Gender differences in joint biomechanics during walking: Normative study in young adults. *American Journal of Physical Medicine and Rehabilitation, 77,* 2-7.

Kerrigan, D. C., Todd, M. K., Della Croce, U., Lipsitz, L. A., & Collins, J. J. (1998). Biomechanical gait alterations independent of speed in the healthy elderly: Evidence for specific limiting impairments. *Archives of Physical Medicine and Rehabilitation, 79,* 317-322.

Kerrigan, D. C., Xenopoulos-Oddsson, A., Sullivan, M. J., Lelas, J. J., & Riley, P. O. (2003). Effect of a hip flexor-stretching program on gait in the elderly. *Arch Phys Med Rehabil, 84,* 1-6.

Kiesel, K., Plisky, P., & Butler, R. (2009). Functional movement test scores improve following a standardized off-season intervention program in professional football players. *Scandinavian Journal of Medicine & Science in Sports,*

Kingma, I., Bosch, T., Bruins, L., & van Dieen, J. H. (2004). Foot positioning instruction, initial vertical load position and lifting technique: Effects on low back loading. *Ergonomics, 47*(13), 1365-1385.

Kingma, I., de Looze, M. P., Toussaint, H. M., Klijnsma, H. G., & Bruijnen, T. B. M. (1996). Validation of a full body 3-D dynamic linked segment model. *Human Movement Science, 15*, 833-860.

Kubo, M., Holt, K. G., Saltzman, E., & Wagenaar, R. C. (2006). Changes in axial stiffness of the trunk as a function of walking speed. *Journal of Biomechanics, 39*, 750-757.

Kujala, U. M., Salminen, J. J., Taimela, S., Oksanen, A., & Jaakkola, L. (1992). Subject characteristics and low back pain in young athletes and nonathletes. *Medicine and Science in Sports and Exercise, 24*(6), 627-632.

Kujala, U. M., Taimela, S., Oksanen, A., & Salminen, J. J. (1997). Lumbar mobility and low back pain during adolescence. *American Journal of Sports Medicine, 25*(3), 363-368.

Kujala, U. M., Taimela, S., Salminen, J. J., & Oksanen, A. (1994). Baseline anthropometry, flexibility and strength characteristics and future low-back pain in adolescent athletes and nonathletes. *Scandinavian Journal of Medicine & Science in Sports, 4*, 200-205.

Kuukkanen, T., & Mälkiä, E. (2000). Effects of a three-month therapeutic exercise program on flexibility in subjects with low back pain. *Physiotherapy Research International, 5*(1), 46-61.

Latash, M. L., & Zatsiorsky, V. M. (1993). Joint stiffness: Myth or reality? *Human Movement Science, 12,* 653-692.

Lea, R. D., & Gerhardt J.J. (1995). *Range of motion measurements*

Lee, L. W., Kerrigan, D. C., & Della Croce, U. (1997). Dynamic implications of hip flexion contractures. *American Journal of Physical Medicine and Rehabilitation, 76,* 502-508.

Lee, R. Y. W., & Wong, T. K. T. (2002). Relationship between the movements of the lumbar spine and hip. *Human Movement Science, 21*, 481-494.

Lephart, S. M., Smoliga, J. M., Myers, J. B., Sell, T. C., & Tsai, Y. S. (2007). An eight-week golf-specific exercise program improves physical characteristics, swing mechanics, and golf performance in recreational golfers. *Journal of Strength and Conditioning Research / National Strength & Conditioning Association, 21*(3), 860-869.

Lieber, R. L., Leonard, M. E., Brown, C. G., & Trestik, C. L. (1991). Frog semitendinosis tendon load-strain and stress-strain properties during passive loading. *American Journal of Physical Medicine*, *261,* C86-C92.

Lu, T. W., Chien, H. L., & Chen, H. L. (2007). Joint loading in the lower extremities during elliptical exercise. *Medicine and Science in Sports and Exercise, 39*(9), 1651-1658.

Maas, H., Jasper, R. T., Baan, G. C., & Huijing, P. A. (2003). Myofascial force transmission between a single muscle head and adjacent tissues: Length effects of head III of rat EDL. *Journal of Applied Physiology, 95*, 2004-2013.

Macnab, I. (1977). *Backache*. Baltimore, Md.: Waverly Press, Inc.

Madding, S. W., Wong, J. G., Hallum, A., & Medeiros, J. M. (1987). Effect of duration of passive stretch on hip abduction range of motion. *The Journal of Orthopaedic and Sports Physical Therapy, 8*, 409-416.

Magee, D. J. (1987). *Orthopedic physical assessment*. Philadelphia, PA: W. B. Saunders Company.

Magnusson, S. P. (1998). Passive properties of human skeletal muscle during stretch maneuvers. *Scandinavian Journal of Medicine and Science in Sports, 8*, 65-77.

Magnusson, S. P., Aagaard, P., Simonsen, E. B., & Bojsen-Moller, F. (2000). Passive tensile stress and energy of the human hamstring muscles *in vivo*. *Scandinavian Journal of Medicine and Science in Sports, 10*, 351-359.

Magnusson, S. P., Simonsen, E. B., Aagaard, P., Gleim, G. W., & Kjaer, M. (1995). Viscoelastic response to repeated static stretching in the human hamstring muscle. *Scandinavian Journal of Medicine & Science in Sports*, *5*, 342-347.

Magnusson, S. P., Simonsen, E. B., Aagaard, P., & Kjaer, M. (1996). Biomechanical responses to repeated stretches in human hamstring muscle in vivo. *The American Journal of Sports Medicine*, *24*(5), 622-628.

Magnusson, S. P., Simonsen, E. B., Aagaard, P., Sorensen, H., & Kjaer, M. (1996). A mechanism for altered flexibility in human skeletal muscle. *Journal of Physiology, 497.1*, 291-298.

Magnusson, S. P., Simonsen, E. B., Dyhre-Poulsen, P., Aagaard, P., Mohr, T., & Kjaer, M. (1996). Viscoelastic stress relaxation during static stretch in human skeletal muscle in the absence of EMG activity. *Scandinavian Journal of Medicine and Science in Sports, 6*, 323-328.

Malliaras, P., Hogan, A., Nawrocki, A., Crossley, K., & Schache, A. (2009). Hip flexibility and strength measures: Reliability and association with athletic groin pain. *British Journal of Sports Medicine, 43*(10), 739-744.

Manning, C., & Hudson, Z. (2009). Comparison of hip joint range of motion in professional youth and senior team footballers with age-matched controls: An indication of early degenerative change? *Physical Therapy in Sport : Official Journal of the Association of Chartered Physiotherapists in Sports Medicine, 10*(1), 25-29.

Marras, W. S., & Granata, K. P. (1997). The development of an EMG-assisted model to assess spine loading during whole-body free-dynamic lifting. *Journal of Electromyography and Kinesiology : Official Journal of the International Society of Electrophysiological Kinesiology, 7*(4), 259-268.

Marras, W. S., Knapik, G. G., & Ferguson, S. (2009). Loading along the lumbar spine as influence by speed, control, load magnitude, and handle height during pushing. *Clinical Biomechanics (Bristol, Avon), 24*(2), 155-163.

Marshall, L. W., & McGill, S. M. (2010). The role of axial torque in disc herniation. *Clinical Biomechanics (Bristol, Avon), 25*(1), 6-9.

Martin, B. W., Sommer, C. J., Ackermann-Liebrich, U., & Baumann, J. U. (1997). Muscle contractures in runners. predictors of range of motion in a cross-sectional study and biomechanical parameters in a small sub-sample. *Gait and Posture, 6,* 272.

McCarthy, J., Noble, P., Aluisio, F. V., Schuck, M., Wright, J., & Lee, J. A. (2003). Anatomy, pathologic features, and treatment of acetabular labral tears. *Clinical Orthopaedics and Related Research, 406*, 38-47.

McClure, P. W., Esola, M. A., Schreier, R., & Siegler, S. (1997). Kinematic analysis of lumbar and hip motion while rising from a forward, flexed position in patients with and without a history of low back pain. *Spine, 22*(5), 552-558.

McCue, B. F. (1953). Flexibility measurements of college women. *Research Quarterly, 24*, 316-324.

McGill, S., Seguin, J., & Bennett, G. (1994). Passive stiffness of the lumbar torso in flexion, extension, lateral bending, and axial rotation. *Spine, 19*(6), 696-670.

McGill, S. M., & Norman, R. W. (1986). Partitioning of the L4-L5 dynamic moment into disc, ligamentous, and muscular components during lifting. *Spine, 11*(7), 666-678.

McGill, S. (2002). *Low back disorders*. Champaign: Human Kinetics.

McGregor, A. H., McCarthy, I. D., & Hughes, S. P. (1995). Motion characteristics of the lumbar spine in the normal population. *Spine, 20*(22), 2421-2428.

Mellin, G. (1988). Correlations of hip mobility with degree of back pain and lumbar spinal mobility in chronic low-back pain patients. *Spine, 13*(1-5), 668-670.

Mellin, G. (1990). Decreased joint and spinal mobility associated with low back pain in young adults. *Journal of Spinal Disorders, 1*(3), 238-243.

Mier, C. M., & Feito, Y. (2006). Metabolic cost of stride rate, resistance, and combined use of arms and legs on the elliptical trainer. *Research Quarterly for Exercise and Sport, 77*(4), 507-513.

Milosavljevic, S., Milburn, P. D., & Knox, B. W. (2005). The influence of occupation on lumbar sagittal motion and posture. *Ergonomics, 48*(6), 657-667.

Möller, M., Ekstrand, J., Öberg, B., & Gillquist, J. (1985). Duration of stretching on range of motion in lower extremities. *Arch Phys Med Rehabil, 66*, 171-173.

Moreside, J. M., & McGill, S. M. (submitted). Passive supine vs prone hip rotation measurement. *Physical Therapy,*

Moreside, J. M., Vera-Garcia, F. J., & McGill, S. M. (2006). Trunk muscle activation patterns, lumbar compressive forces, and spine stability when using the bodyblade. *Physical Therapy, 87*(2), 153-163.

Mundale, M. O., Hislop, H. J., Rabideau, R. J., & Kottke, F. J. (1956). Evaluation of extension of the hip. *Archives of Physical Medicine and Rehabilitation, 37*, 75-80.

Murray, M. P. (1967). Gait as a total pattern of movement. *American Journal of Physical Medicine, 46*(1), 290-333.

Myers, T. W. (2001). *Anatomy trains: Myofascial meridians for manual and movement therapists*. New York: Churchill Livingstone.

Nachemson, A. (1966). The load on lumbar disks in different positions of the body. *Clinical Orthopaedics and Related Research, 45*, 107-122.

Nachemson, A. L., & Morris, J. M. (1964). In vivo measurements of intradiscal pressure. *Journal of Bone and Joint Surgery, 46A*, 1077-1092.

Norman, R., Wells, R., Neumann, P., Frank, J., Shannon, H., & Kerr, M. (1998). A comparison of peak vs cumulative physical work exposure risk factors for the reporting of low back pain in the automotive industry. *Clinical Biomechanics (Bristol, Avon), 13*(8), 561-573.

Nourrit, D., Delignieres, D., Caillou, N., Deschamps, T., & Lauriot, B. (2003). On discontinuities in motor learning: A longitudinal study of complex skill acquisition on a ski-simulator. *Journal of Motor Behavior, 35*(2), 151-170.

Offierski, C. M., & Macnab, I. (1983). Hip-spine syndrome. *Spine, 8*(3), 316-321.

Öhlen, G., Wredmark, T., & Spangford, E. (1989). Spinal sagittal configuration and mobility related to low-back pain in the female gymnast. *Spine*, *14*(-), 847-850.

Olson, M., Solomonow, M., & Li, L. (2006). Flexion-relaxation response to gravity. *Journal of Biomechanics, 39*(14), 2545-2554.

Orchard, J. W., Farhart, P., & Leopold, C. (2004). Lumbar spine region pathology and hamstring and calf injuries in athletes: Is there a connection? *British Journal of Sports Medicine, 38*, 502-504.

Panjabi, M. M. (2003). Clinical spinal instability and low back pain. *Journal of Electromyography and Kinesiology, 13*, 371-379.

Paquet, N., Malouin, F., & Richards, C. L. (1994). Hip-spine movement interaction and muscle activation patterns during sagittal trunk movements in low back pain patients. *Spine, 19*(5), 596-603.

Parkinson, R. J., Beach, T. A. C., & Callaghan, J. P. (2004). The time-varying resonse of the in vivo lumbar spine to dynamic repetitive flexion. *Clinical Biomechanics, 19*, 330-336.

Pearcy, M. J., Portek, I., & Shepherd, J. T. (1984). Three-dimensional X-ray analysis of normal movement in the lumbar spine. *Spine, 9*(3), 294-297.

Pearcy, M. J., & Tibrewal, S. B. (1984). Axial rotation and lateral bending in the normal lumbar spine measured by three-dimensional radiography. *Spine*, *9*(6), 582-587.

Pearcy, M. J., & Whittle, M. W. (1982). Movements of the lumbar spine measured by three-dimensional x-ray analysis. *Journal of Biomedical Engineering, 4*, 107-112.

Perry, J. (1992). *Gait analysis*. Thorofare, NJ: SLACK Inc.

Pollard, C. D., Sigward, S. M., & Powers, C. M. (2007). Gender differences in hip joint kinematics and kinetics during side-step cutting maneuver. *Clinical Journal of Sport Medicine, 17*, 38-42.

Prentice, W. E. (1983). A comparison of static stretching and PNF stretching for improving hip joint flexibility. *Athletic Training, 1983*(18), 56-59.

Rijkelijkhuizen, J. M., Baan, G. C., de Haan, A., de Ruiter, C. J., & Huijing, P. A. (2005). Extramuscular myofascial force transmission for *in situ* rat medial gastrocnemius and plantaris muscles in progressive stages of dissection. *The Journal of Experimental Biology, 208*, 129-140.

Riley, P. O., Della Croce, U., & Kerrigan, D. C. (2001). Effect of age on lower extremity joint moment contributions to gait speed. *Gait and Posture, 14*, 264-270.

Riley, P. O., Dicharry, J., Franz, J., Della Croce, U., Wilder, R. P., & Kerrigan, D. C. (2008). A kinematics and kinetic comparison of overground and treadmill running. *Medicine and Science in Sports and Exercise, 40*(6), 1093-1100.

Roach, K. E., & Miles, T. P. (1991). Normal hip and knee active range of motion: The relationship to age. *Physical Therapy, 71*(9), 656-665.

Roberts, J. M., & Wilson, K. (1993). Effect of stretching duration on active and passive range of motion in the lower extremity. *British Journal of Sports Medicine, 33*, 259-263.

Ross, M. D., Nordeen, M. H., & Barido, M. (2003). Test-retest reliability of patrick's hip range of motion test in healthy college-aged men. *Journal of Strength and Conditioning Research / National Strength & Conditioning Association, 17*(1), 156-161.

Rowe, P. J., & White, M. (1996). Three dimensional, lumbar spinal kinematics during gait, following mild musculo-skeletal low back pain in nurses. *Gait and Posture, 4*, 242-251.

Sady, S. P., Wortmann, M., & Blanke, D. (1982). Flexibility training: Ballistic, static or proprioceptive neuromuscular proprioception? *Arch Phys Med Rehabil, 63*, 261-263.

Sahrmann, S. (2002). *Diagnosis and treatment of muscle imbalances associated with regional pain syndromes; a course manual*. St. Louis: Mosby.

Scannell, J., & McGill, S. (2003). Lumbar posture - should it, and can it, be modified? A study of passive tissue stiffness and lumbar position during activities of daily living. *Physical Therapy, 83*(10), 907-917.

Schache, A. G., Blanch, P. D., & Murphy, A. T. (2000). Relation of anterior pelvic tilt during running to clinical and kinematic measures of hip extension. *British Journal of Sports Medicine, 34*, 279-283.

Schache, A. G., Blanch, P. D., Rath, D. A., Wrigley, R. V., & Bennell, K. L. (2005). Are anthropometric and kinematic parameters of the lumbo-pelvic complex related to running injuries? *Research in Sports Medicine*, *13*, 127-147.

Schache, A. G., Blanch, P. D., Rath, D. A., Wrigley, T., & Bennell, K. L. (2003). Differences between the sexes in the three-dimensional angular rotations of the lumbo-pelvic-hip complex during treadmill running. *Journal of Sports Science, 21*, 105-118.

Schache, A. G., Blanch, P. D., Rath, D. A., Wrigley, T. V., Starr, R., & Bennell, K. L. (2002). Intra-subject repeatability of the three dimensional angular kinematics within the lumbo-pelvic-hip complex during running. *Gait and Posture, 15*, 136-145.

Scholtes, S. A., Gombatto, S. P., & Van Dillen, L. R. (2009). Differences in lumbopelvic motion between people with and people without low back pain during two lower limb movement tests. *Clinical Biomechanics (Bristol, Avon), 24*(1), 7-12.

Schulitz, K. P., & Niethard, F. U. (1980). Strain on the interarticular stress distribution. measurements regarding the development of spondylolysis. *Archives of Orthopaedic Trauma Surgery, 96*, 197-202.

Shirazi-Adl, A. (1991). Finite-element evaluation of contact loads on facets of an L2-L3 lumbar segment in complex loads. *Spine, 16*(-), 533-541.

Shirazi-Adl, A. (1994). Biomechanics of the lumbar spine in sagittal/lateral moments. *Spine, 19*(21), 2407-2414.

Shirazi-Adl, A., & Drouin, G. (1987). Load-bearing role of facets in a lumbar segment under sagittal plane loadings. *Journal of Biomechanics, 20*(6), 601-613.

Shum, G. L. K., Crosbie, J., & Lee, R. Y. W. (2005). Symptomatic and asymptomatic movement coordination of the lumbar spine and hip during an everyday activity. *Spine*, *30*(23), E697-E702.

Shum, G. L. K., Crosbie, J., & Lee, R. Y. W. (2007). Movement coordination of the lumbar spine and hip during a picking up activity in low back pain subjects. *Eur Spine J, 16,* 749-758.

Simoneau, G. G., Hoenig, K. J., Lepley, J. E., & Papanek, P. E. (1998). Influence of hip position and gender on active hip internal and external rotation. *Journal of Orthopaedic and Sports Physical Therapy, 28*(3), 158-164.

Sjolie, A. N. (2004). Low back pain in adolescents is associated with poor hip mobility and high body mass index. *Scandinavian Journal of Medicine & Science in Sports, 14*, 168-175.

Sjölie, A. N., & Ljunggren, A. E. (2001). The significance of high lumbar mobility and low lumbar strength for current and future low back pain in adolescents. *Spine*, *26*(23), 2629-2636.

Slauterbeck, J., Clevenger, C., Lundberg, W., & Burchfield, D. M. (1999). Estrogen level alters the failure load of the rabbit anterior cruciate ligament. *Journal of Orthopaedic Research : Official Publication of the Orthopaedic Research Society*, *17*(3), 405-408.

Slauterbeck, J. R., Fuzie, S. F., Smith, M. P., Clark, R. J., Xu, K., Starch, D. W., et al. (2002). The menstrual cycle, sex hormones, and anterior cruciate ligament injury. *Journal of Athletic Training, 37*(3), 275-278.

Smidt, G. L., McQuade, K., Wei, S., & Barakatt, E. (1995). Sacroiliac kinematics for reciprocal straddle positions. *Spine, 20*(9), 1047-1054.

Solomonow, M., Baratta, R. V., Banks, A., Freudenberger, C., & Zhou, B. H. (2003). Flexion-relaxation response to static lumbar flexion in males and females. *Clinical Biomechanics, 18*, 273-279.

Stecco, A., Macchi, V., Masiero, S., Porzionato, A., Tiengo, C., Stecco, C., et al. (2009). Pectoral and femoral fasciae: Common aspects and regional specializations. *Surgical and Radiologic Anatomy : SRA, 31*(1), 35-42.

Stecco, C., Porzionato, A., Macchi, V., Tiengo, C., Parenti, A., Aldegheri, R., et al. (2006). Histological characteristics of the deep fascia of the upper limb. *Italian Journal of Anatomy and Embryology = Archivio Italiano Di Anatomia Ed Embriologia, 111*(2), 105-110.

Sullivan, M. S., Dickinson, C. E., & Troup, J. D. (1994). The influence of age and gender on lumbar spine sagittal plane range of motion. A study of 1126 healthy subjects. *Spine, 19*(6), 682-686.

Tampier, C., Drake, J. D., Callaghan, J. P., & McGill, S. M. (2007). Progressive disc herniation: An investigation of the mechanism using radiologic, histochemical, and microscopic dissection techniques on a porcine model. *Spine, 32*(25), 2869-2874.

Taylor, D. C., Dalton, J. D., Seaber, A. V., & Garrett, W. E. J. (1990). Viscoelastic properties of muscle-tendon units. *The American Journal of Sports Medicine, 18*(3), 300-309.

Taylor, J., & Twomey, L. T. (1980). Sagittal and horizontal plane movement of the human lumbar vertebral column in cadavers and in the living. *Rheumatology and Rehabilitation, 19*, 223-232.

Taylor, J. R., & Twomey, L. T. (1986). Age changes in lumbar zygapophyseal joints. *Spine, 11*, 739-745.

Thacker, S. B., Gilchrist, J., Stroup, D. F., & Kimsey, C. D. J. (2004). The impact of stretching on sports injury risk: A systematic review of the literature. *Med Sci Sports Exercise, 36*(3), 371-378.

Thurston, A. J. (1985). Spinal and pelvic kinematics in osteoarthrosis of the hip joint. *Spine, 10*(1-5), 467-471.

Thurston, A. J., & Harris, J. D. (1983). Normal kinematics of the lumbar spine and pelvis. *Spine, 8*(2), 199-205.

Ticker, J. B., Bigliani, L. U., Soslowsky, L. J., Pawluk, R. J., Flatow, E. L., & Mow, V. C. (1996). Inferior glenohumeral ligament: Geometric and strain-rate dependent properties.

Journal of Shoulder and Elbow Surgery / American Shoulder and Elbow Surgeons ...[Et Al.], 5(4), 269-279.

Torry, M. R., Schenker, M. L., Martin, H. D., Hogoboom, B. S., & Philippon, M. J. (2006). Neuromuscular hip biomechanics and pathology in the athlete. *Clinics in Sports Medicine, 25*, 179-197.

Tsai, L., & Wredmark, T. (1993). Spinal posture, sagittal mobility, and subjective rating of back problems in former female elite gymnasts. *Spine, 18*(7), 872-875.

Twomey, L. T. (1983). Sagittal movements of the human lumbar vertebral column: A quantitative study of the role of the posterior vertebral elements. *Arch Phys Med Rehabil, 64*, 322-325.

Vad, V. B., Bhat, A., Basrai, D., Gebeh, A., Aspergren, D. D., & Andrews, J. R. (2004). Low back pain in professional golfers; the role of associated hip and low back range-of-motion deficits. *The American Journal of Sports Medicine, 32*(2), 494-497.

Vad, V. B., Gebeh, A., Dines, D., Altchek, D., & Norris, B. (2003). Hip and shoulder internal rotation range of motion deficits in professional tennis players. *Journal of Science and Medicine in Sport, 6*(1), 71-75.

Van Dillen, L. R., Bloom, N. J., Gombatto, S. P., & Susco, T. M. (2008). Hip rotation range of motion in people with and without low back pain who participate in rotation-related sports. *Physical Therapy in Sport : Official Journal of the Association of Chartered Physiotherapists in Sports Medicine, 9*(2), 72-81.

Van Dillen, L. R., Gombatto, S. P., Collins, D. R., Engsberg, J. R., & Sahrmann, S. (2007). Synmmetry of timing of hip and lumbopelvic rotation motion in 2 different subgroups of people with low back pain. *Arch Phys Med Rehabil, 88,* 351-360.

Van Dillen, L. R., McDonnell, M. K., Fleming, D. A., & Sahrmann, S. (2000). Effect of knee and hip position on hip extension range of motion in individuals with and without low back pain. *Journal of Orthopaedic and Sports Physical Therapy, 30*(6), 307-316.

Vereijken, B., van Emmerik, F. A. E., Bonnefoy, R., Beek, W. J., & Newell, K. M. (1997). Changing coordinative structures in complex skill acquisition. *Human Movement Science, 16,* 823-844.

Verrall G.M., Slavotinek J.P., Barnes P.G., Esterman A., Oakeshott R.D., & Spriggins A.J. (2007). Hip joint range of motion restriction precedes athletic chronic groin injury. *Journal of Science and Medicine in Sport, ,* e-version.

Wallin, D., Ekblom, B., Grahn, R., & Nordenborg, T. (1985). Improvement of muscle flexibility. *The American Journal of Sports Medicine*, *13*(4), 263-268.

Weldon, S. M., & Hill, R. H. (2003). The efficacy of stretching for prevention of exercise-related injury: A systematic review of the literature. *Manual Therapy*, *8*(3), 141-150.

Whittle, M. W., & Levine, D. F. (1995). Sagittal plane motion of the lumbar spine during normal gait. *Gait and Posture, 3,* 82.

Williams K.R., Cavanagh, P. R., & Ziff, J. L. (1987). Biomechanical studies of elite female distance runners. *International Journal of Sports Medicine, 8*, 107-118.

Williams, P. E. (1990). Use of intermittent stretch in the prevention of serial sarcomere loss in immobilised muscle. *Annals of the Rheumatic Diseases, 49*, 316-317.

Williams, P. E., & Goldspink, G. (1973). The effect of immobilization on the longitudinal growth of striated muscle fibres. *Journal of Anatomy, 116*(1), 45-55.

Winters, M. V., Blake, C. G., Trost, J. S., Marcello-Brinker, T. B., Lowe, L., Garber, M., et al. (2004). Passive versus active stretching of hip flexor muscles in subjects with limited hip extension: A randomized clinical trial. *Physical Therapy, 84,* 800-807.

Wojtys, E. M., Huston, L. J., Boynton, M. D., Spindler, K. P., & Lindenfeld, T. N. (2002). The effect of the menstrual cycle on anterior cruciate ligament injuries in women as determined by hormone levels. *The American Journal of Sports Medicine, 30*(2), 182-188.

Wong, T. K. T., & Lee, R. Y. W. (2004). Effects of low back pain on the relationship between the movements of the lumbar spine and hip. *Human Movement Science, 23*, 21-34.

Zajac, F. E., Neptune, R. R., & Kautz, S. A. (2003). Biomechanics and muscle coordination of human walking. part II: Lessons from dynamical simulations and clinical implications. *Gait and Posture, 17*, 1-17.