BIOMECHANICAL COMPARISONS CONSIDERING RISK TO THE LUMBAR SPINE: WALKING WITH NO LOAD, A BACKPACK, AND A PERSON ON THE BACK

By:

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ABSTRACT

The purpose of this study was to determine biomechanical differences between carrying no load, a backpack, and a person on the back, in relation to risk to the lumbar spine. The common piggyback carry has recently become more popular through the trending sport of CrossFit. No previous research literature on the piggyback carry has been located, but some biomechanics research exists on the similar activity of backpack carriage.

Twelve 70+ kg male strength-trained athletes that regularly lift at least 40 kg and had no recent musculoskeletal injuries were recruited from local CrossFit affiliates. One child with a mass of 29 kg was recruited to be the piggyback passenger for all participants. All participants and the guardian of the passenger signed an informed consent form. The participants walked three times over a force plate embedded in an eight metre walkway for each of three conditions: carrying no load, a 29 kg backpack, or a 29 kg passenger. The unloaded condition was performed first for all participants; the subsequent order for the loaded conditions was random. Data collection required that the right foot make full contact with the force plate, and only the force plate, while walking naturally for a trial to be considered successful.

Three lateral view Canon video cameras recorded each trial: left and right views and a zoomed-in right view, focussed on the foot placement. Dartfish Software was used to measure joint angles and gait parameters. Maximal trunk flexion, hip flexion, knee flexion angles and trunk range of motion were measured, as well as external torques, and velocity, cadence, and stance, single stance, and double stance durations. Single stance duration was also measured using the AMTI force plate. Maximal trunk flexion angle, the key variable in this study, was used in a static lumbar spine model to calculate estimations of trunk muscle torque and force, and compressive, shear and resultant lumbosacral joint reaction forces. The force plate measured vertical ground reaction force (vGRF) data. The vGRFs were graphed over time, and three specific magnitudes were measured: the first maximal peak, the interpeak dip, and second maximal peak of the vGRF.

All variables were compared using a repeated measures ANOVA test with a Bonferroni correction. Both load conditions produced compensatory trunk flexion; trunk flexion increased from no load to piggybacking to backpacking. Trunk range of motion was similar for no load and piggybacking, but increased to backpacking. Hip angle changes did not show significance, but maximal knee flexion angle was greater for backpacking than the other conditions. The backpack load caused greater resultant and total magnitude of torque than the passenger load. The total magnitude of external torque and resultant joint reaction forces increased from no load to piggybacking to backpacking. The trunk extensors dominated with no load and piggybacking and the trunk flexors dominated with backpacking.

Many of the significant differences between conditions suggest that piggybacking is biomechanically more similar to natural gait than is backpacking. Due to the more extreme position of the trunk, greater trunk range of motion, and greater magnitude of external torques, backpacking likely places the musculoskeletal system at more risk than does piggybacking.

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CHAPTER I

INTRODUCTION

GENERAL OVERVIEW

The piggyback is a unique recreational and training skill that has not been studied nor scientifically questioned, yet it is a relatively common activity and may pose a risk to the lumbar spine of individuals whose musculoskeletal systems are not yet adapted to carry the magnitude of load of the person that they are carrying. Anyone may perform this activity and, unlike most loaded activities for which lumbar forces have been estimated, the load is carried primarily posterior to the trunk. During the piggyback, the carrier must support the weight of both the passenger and him or herself, while maintaining balance and moving forward. Carrying load, and specifically carrying a load posterior to the spine, has been identified as an activity that should be considered for possible risk to the lumbar spine by lifting studies (Cholewicki, McGill, & Norman, 1991) and backpack studies (Chow, Kwok, Au-yang, Holmes, Cheng, Yao, & Wong, 2005; Goh, Thambyah, & Bose, 1998). Addressing the trunk flexion angle and estimating the compressive, shear, and resultant joint reaction forces at the lumbosacral joint during this activity will help to address such risk.

Most Canadians have likely either been carried by another person, have carried another person, or have experienced both roles in the execution of a piggyback, as seen in Figure 1.1. The piggyback is not only an important part of family life and social culture, as a common bonding and rough-housing experience important to social and physical development within a family or friendship (Bennett, 1999), but it is also used by athletes for strength training. This skill has been used for resistance training by boot camps, soccer teams and rugby teams

(Sportsmail Reporter, 2011; Nottingham Rugby, 2011; Piggyback New West, 2013), and has become a larger part of the athletic community through the recent trend of CrossFit (CrossFit Inc., 2008).

Figure 1.1: The piggyback carry from sagittal, anterior and posterior views.

With the realization of an obesity epidemic, bootcamps and other types of innovative fitness programs have become more popular within the last couple decades. This eventually inspired the development of the individual and team sport of CrossFit, which utilizes the piggyback carry in training and competition. CrossFit is a method of physical training and strength development with its own set of philosophies and competitions, and integrates exercises and physical challenges from a variety of disciplines (The Box Magazine, 2012). CrossFit was developed by teenage gymnast and cyclist Greg Glassman in the years before he formally established his company in 2000 (The Box Magazine, 2012), and it has grown into a sport over the past several years. The first affiliate was CrossFit North in Seattle in 2000 and by 2012, there were 3400 affiliates around the world (Sternkopf, 2012). There were six CrossFit Fitness centres with an internet presence in Winnipeg in the spring of 2013; this number grew to eleven as of December 2014.

During CrossFit training, athletes perform a variety of exercises that could be described as a combination of activities of daily living and resistance training, generally with the goal of surpassing previous self-perceived limits of physical capability. Examples of exercises are walking lunges, pull-ups, tire lifts, the kettle bell snatch (CrossFit Winnipeg, 2014), and variations of the piggyback (CrossFit Inc., 2008; CrossFit Winnipeg, 2013). The CrossFit piggyback may be performed while standing, walking, or during other activities such as calf raises or squats to add load for more resistance. The piggyback was also included in a timed circuit repeated five times that included various exercises and a 50 yard piggyback dash (CrossFit, 2014). The creator of this workout of the day (WOD) allowed a loaded backpack to be substituted for the person during the piggyback portion of the challenge (CrossFit Inc., 2014).

Description of the Skill

During the piggyback, the body of the passenger is posterior to and wrapped around the carrier as seen in Figure 1; the arms wrap over the shoulders and anteriorly around the neck while the legs wrap laterally and are held by the carrier superior to the iliac crest. The weight of the passenger is supported partly by the carrier exerting a lifting force on the legs of the passenger, the arms of the passenger exerting a downward force on the shoulder girdle of the carrier, and the friction and weight of the passenger's pelvis resting slightly on the lower back of the carrier. The passenger can help support more of his or her weight by exerting a greater force on the shoulder girdle of the carrier, but the entire weight of the passenger is basically supported by the carrier superior to the lumbosacral joint. For the purposes of this study, the trunk will be considered to be in an approximately constant position of flexion during the piggyback while the lower body transports it across the ground.

Typically known as the piggyback, it is also known as the partner carry (Muller, 2013) or buddy carry (BC Crossfit Brier Creek, 2013), although either of the two latter terms can also refer to a position known as the fireman's carry. CrossFit athletes may also carry a person in a position such as a firemans carry (CrossFit Winnipeg, 2013), or other positions that have been seen in athletic training or competition as seen in Figure 1.2. The person may be carried more superiorly with the legs and arms draped over opposite shoulders, known as the fireman's carry, may sit on the carrier's shoulders, known as the shoulder sit, or the feet may be draped over the shoulders anteriorly at the knee with the trunk posterior to that of the carrier, known as the wife carry. Although there are several positions in which to carry another person, this thesis will solely address the kinetics and kinematics of the piggyback carry.

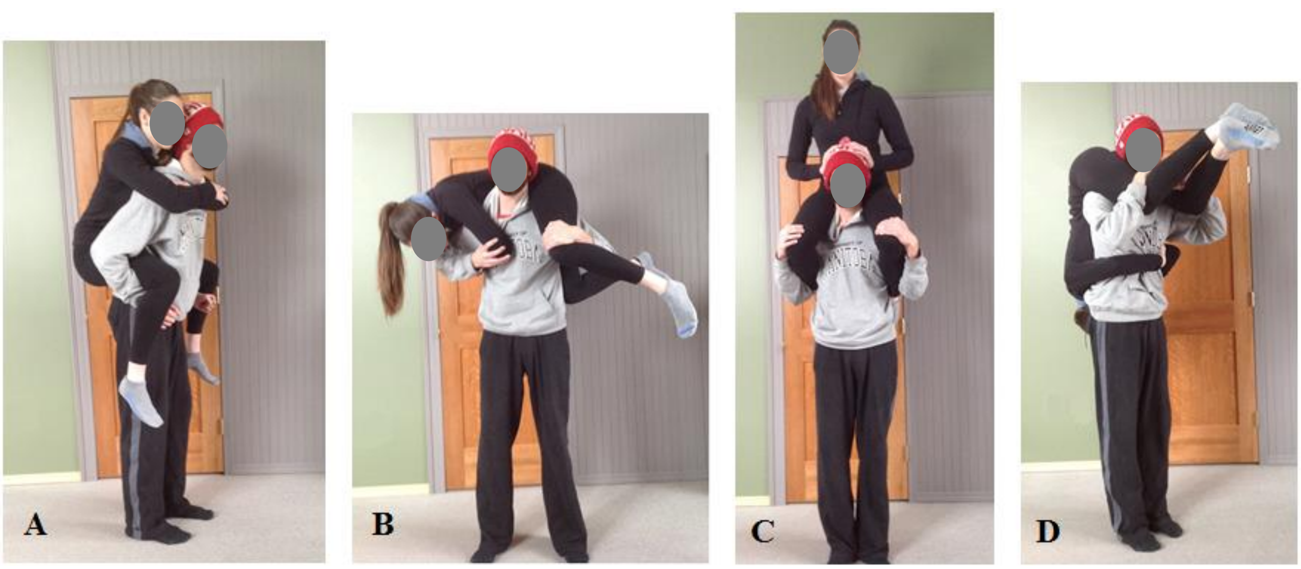


Figure 1.2: Methods of carrying another person. A) the piggyback, B) the fireman's carry, C) the shoulder sit, and D) the wife carry.

RATIONALE FOR THE STUDY

Individuals commonly piggyback other individuals in every day settings, during training and conditioning, and more recently in the sport of CrossFit, loading the back while back health is a prominent issue in society. Piggyback carriage should be biomechanically investigated because:

1) The lumbar spine is vulnerable and injury to this area is prevalent.

2) The piggyback is a common public activity about which little is known biomechanically. It is an interesting activity involving balance, strength, coordination, an active passenger, and locomotion.

3) It has been increasingly popular as a training tool for CrossFit and other fitness groups, and as a challenge in competitions such as the CrossFit Games.

Prevalence of Lumbar Back Pain

Pain or injury of the back is one of the most common chronic conditions in the Canadian population (Schultz & Kopec, 2003), and it therefore substantially impacts both the health care system and the workforce. Approximately 80 % of the population will endure at least one episode of back pain in their lifetime (Statistics Canada, 2006). From 2008 to 2012, back strain was the single most common injury resulting in time loss claims by young workers in British Columbia (WorkSafeBC, 2013), and low back pain specifically was the second greatest reason for missing work in the United States (Hall, 2007). Approximately70% of American work time lost because of back injury was due to issues of the lumbar spine (Hall, 2007). In fact, most of the cases involving pain or injury to the back involve the lumbar region (Hall, 2007), and most disc injuries occur at the lumbosacral joint (Hall, 2007; Khoo et al., 1995). Regular piggybacking could be placed in the activity category involving frequent weight-loading, which is one of the three types of activity categories that most commonly injure the back (Alexander, 1985).

Because the lumbar spine is known to be a common site of injury, there is a significant amount of literature on the demands on the lumbar spine and lumbar kinetics such as net moments, joint reaction forces, and intervertebral disc compression (McGill & Norman, 1985). Compressive stress on the low back can lead to end plate fracture, herniated discs, and sciatica (McGill & Norman, 1985). The compressive force on the lumbosacral disc has been a large influence on suggested load limits, such as the NIOSH equation which has been used in the workplace for decades (1981)(McGill & Norman, 1985). Previous research has resulted in the NIOSH equation, a calculation used as a guideline for workplace load maximums. The 1993 NIOSH recommendation was a maximum disc compressive force of 3400 N (Waters, Putz-Anderson, Garg, & Fine, 1993).

An Interesting and Physically Demanding Activity

Piggybacking is a common technique by which to carry a person, and has somewhat similar kinematics and kinetics to the better-understood activity of carrying a backpack because they are both predominantly posterior loads. Both loads need to be balanced while locomotion is performed, but the piggyback load is unique due to the nature of a live load that is helping to hold itself in place. Strength and coordination are also necessary, and the passenger is often even as heavy as or heavier than the carrier. The prevalence of injuries due to piggybacking is not known, nor is much scientifically known about this activity. Therefore, the mechanics of carrying a person was investigated and compared with the mechanics of carrying a backpack. This research was the first known biomechanical study on the piggyback carry; it can be a stepping stone for further biomechanical exploration of carrying a piggyback load while walking, through the means of kinetic and kinematic analysis.

CrossFit and Training with Safe Progressions

Some CrossFit centres likely use the piggyback as a training tool because people enjoy team bonding and they need to prepare for the activities that they may be asked to perform at the CrossFit Games. Physical training is based on the idea that muscles and soft tissues adapt to loading to become stronger, but it is important for the stress on the tissue be within safe limits. The principle of progressive overload states that too great of an increase in training load too quickly can cause injury and immune disruptions; resistance should only be increased by approximately five percent when able to lift more than twelve repetitions in two consecutive workouts (Fahey, Insel, & Roth, 2007). When an individual carries a heavily loaded backpack or another person on the back, the stress on the lumbosacral joint, or other joints, may exceed safe amounts and increase the risk of injury. If a CrossFit group or a team uses this exercise as a training exercise, at least one athlete will have a partner of a weight greater than their own on their back. This means that possibly without previous graded resistance training, the lightest athlete in the group is suddenly supporting more than double his or her body weight. If an individual regularly carries or lifts other heavy loads, then this load may not pose any extra risk because the body has adapted to such strength levels. However, if the individual is just starting out or has not trained through lifting progressively heavier loads, the appropriate training that would safely lead to this progression may be lacking. These increases in force theoretically place increased stress on the lumbar spine and may lead to injury.

Currently, there are no piggyback guidelines for coaches or athletes to follow, although they are being asked to and expected to administer this drill. They use common sense and any general resistance training knowledge that they have as a guideline when pairing piggybacking partners or considering their form. When bootcamps or teams use the piggyback carry as a drill or when parents use this activity with their children, it would be beneficial for them to understand the forces experienced and demands it places on the lumbar spine. If the biomechanical demands are understood, then appropriate postural, progression, and other guidelines could be established. Any contraindications could be addressed and individuals could be educated on avoiding risk. The results of this research provide insight into whether an individual should train with a backpack or human load first or at all. Logical and safe progressions can now be considered after producing some initial data comparing piggyback to backpack carriage.

Piggybacking is Comparable to Backpacking

Backpacking is biomechanically similar to piggybacking. A load is carried in a location posterior to the trunk of the carrier and the weight is supported by the shoulders, back, and pelvis. The centre of gravity, which is a point or location where an equal amount of mass of the object is on each opposing side of that point, should be considered when addressing the biomechanical similarities and differences. The centre of gravity of the backpack is posterior to the trunk, at some point within the backpack, and therefore causes a backward torque on the trunk of the carrier. A torque is a rotational tendency, caused by a force acting at some distance from an axis of rotation, as seen in Equation 1. Here, the moment arm ($d_\perp$) is the perpendicular

Equation 1: $$T = F\, d_\perp$$

where T represents the torque created by a force acting at some distance from an axis of rotation, F represents the force, and $d_\perp$ represents the moment arm of the force

(Hall, 2007; Levangie & Norkin, 2005; Nordin & Frankel, 2012) distance between the mediolateral axis of rotation in the lumbar spine and the vertical line of the force of gravity on the mass in question.

The centre of gravity of a person being piggybacked is addressed by dividing it into the centre of gravity of each body segment, some segments posterior to the trunk of the carrier and creating a backward rotating torque, but some segments anterior to the trunk of the carrier and creating a forward rotating torque. Therefore, the overall centre of gravity of the passenger is more anterior to the centre of gravity of the backpack, which is the proposed primary biomechanical difference between the two activities as seen in Figures 1.3 and 1.4. Otherwise, both activities load the upper body of the carrier and have some amount of posterior load. Because there was no current literature on the piggyback, the results of a small pilot study performed in June 2013 (Graham) were considered in conjunction with existing literature on the backpack and load in general to hypothesize the possible effects of piggybacking on lumbar forces and gait patterns. Backpack studies have been executed with regard to recreational hiking

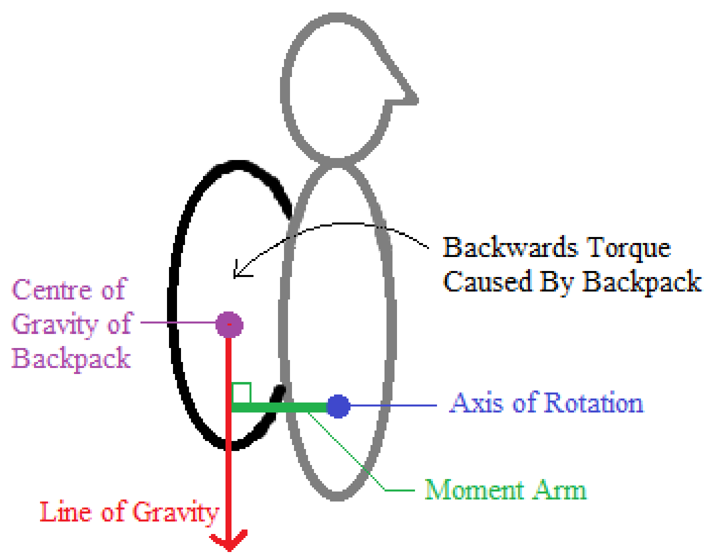


Figure 1.3: Torque caused by a backpack. The force of gravity acting on the centre of gravity of the posterior load creates a tendency for the trunk to rotate backwards around the axis of rotation in the lumbar spine.

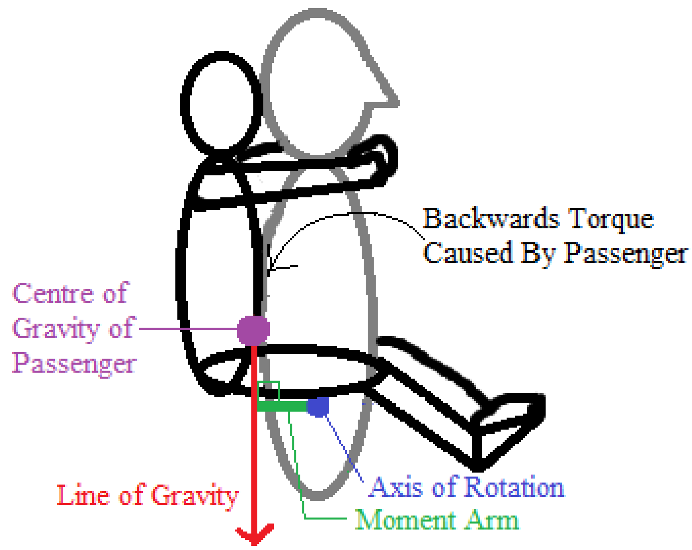


Figure 1.4: Torque caused by piggyback. The force of gravity acting on the centre of gravity of the posterior load creates a tendency for the trunk to rotate backwards around the axis of rotation of the lumbar spine. Note that with a shorter moment arm, less torque will be produced.

packs, military packs, and school bags.

Posterior Load Induces Trunk Flexion

The location of the overall centre of gravity of a person and the load being carried by that person affect the stability of the system. This is determined by the line of gravity and the base of support locations with respect to each other. The line of gravity is a line from the centre of gravity of an object towards the centre of the Earth, representing the force of gravity on that object or segment. The base of support is the area between all points of contact of the person on the ground. For a stationary person to be stable, their line of gravity must be located within their base of support. When a person is wearing a backpack or carrying a person on the back, which likely both initially shift the centre of gravity of the system posteriorly, the carrier may have to alter his or her trunk or limb position in order to ensure that the line of gravity does not move out of the back edge of the base of support. If the line of gravity was posterior to the posterior border of the base of support, the system would not be balanced and the person and the load would tend to fall backwards. It is likely that the carrier will increase flexion of the trunk to shift the centre of gravity of the system forward, closer to the centre of the base of support. This concept is illustrated in Figure 2.20 on page 76 of Chapter II.

It has already been seen that in both the piggybacking and backpacking situations, a posterior load resulted in flexion of the trunk (Birrell, Hooper, & Haslam, 2006; Chow et al., 2005; Goh et al., 1998; Graham, 2013), which Goh et al. (1998) suggested occurred to place the new centre of gravity of the system, including the carrier and the load, over the base of support. Because the centre of gravity of a carried backpack is likely more posterior than the centre of gravity of a carried person, it was expected that the carrier would experience greater trunk flexion, in order to shift the centre of gravity a greater distance forward, when carrying a backpack as compared to a person.

As the upper body moves further into trunk flexion from a neutral position, it produces a greater forward moment. This forward-rotating torque may be balanced by the backward-rotating torque caused by the load, and any difference in these torques is balanced by trunk extensor or flexor muscle torque. As the muscle forces increase, they also increase the stress exerted on their attachment points and cause more compression at the intervertebral joints. The greater the load and the greater the muscle forces necessary to hold the body in this inclined position, the higher the magnitude of the compressive and shear joint reaction forces in the lumbar spine.

Risk to the Lumbar Spine

Load carriage can expose the lumbar spine to large amounts of stress when it is performed with improper posture or excessive load. Because lumbar spine pain or injury may be a result of backpack carriage (Atreya, Johsi, Anand, Singh, & Ribeiro, 2010; Chow et al., 2005;

Cottalorda, Rahmani, Diop, Gautheron, Ebermeyer, & Belli, 2003) and piggybacking is very similar to backpacking, piggybacking was investigated for probable risk to the lumbar spine. Lumbar intervertebral disc compression has been considered a reflection of stress on the low back (McGill & Norman, 1985), but load torques were able to be more accurately measured. Therefore, torques and loads at the lumbosacral joint were estimated and compared between unloaded, piggyback and backpack conditions. The magnitude of load at the lumbar spine depended on the weight of the object, position of the object, and the angle of spinal flexion (McGill & Norman, 1985).

Lumbosacral Joint – The Mediolateral Axis of Rotation

Within the lumbar spine, the lumbosacral intervertebral disc tends to be the most vulnerable (Hall, 2007; Khoo, Goh, & Bose, 1995) and, likely because of this, it was the joint of interest in many studies that predicted forces experienced at the lumbar spine (Goh et al., 1998). It supports the greatest proportion of load, allows a modest amount of mobility (Moore, Dalley, & Agur, 2010), and the resting surface is at a greater angle than for any other intervertebral disc (Levangie & Norkin, 2005; Nordin & Frankel, 2012). The mediolateral axis of rotation about which trunk flexion and extension occurred was considered to be at the lumbosacral joint in the present study.

Lumbar Spine Model

To estimate the stress experienced by the lower back, a mechanical model based on previous lumbar lifting models was used. Static models are commonly used if there are little to no accelerations; a more complicated dynamic model can be used if there are greater accelerations (McGill & Norman, 1985). For simplification and because the back stays near the same position of incline while walking with a load (Goh et al., 1998), a static analysis was utilized. Therefore, equilibrium of the system was considered to be maintained and unknown values were determined by balancing equations. This model followed three important steps: the moments were balanced about the lumbosacral joint and then the forces were balanced in each of the y and the x axes. The y axis was set perpendicular to- and the x axis was set parallel to- the surface of the sacrum, which was assumed to be at a 30 degree angle to the horizontal (Levangie & Norkin, 2005; Nordin & Frankel, 2012). Rotation about the lumbosacral joint in the direction of trunk flexion was considered positive. These equations are explained in full starting on page 16 of Chapter III. Estimated values were compared between conditions, for the consideration of any injury risk, and for possible recommendations for use of the piggyback.

PURPOSE OF THE STUDY

Carrying loads has been known to place stress on the lumbar spine. However, there is no literature to suggest that the biomechanical effects of load have ever been investigated for the piggyback. The expectations and hypotheses were based on brief pilot studies and research that involved load, usually in the form of a backpack. The purpose of this study was to establish a basic idea of the biomechanics of piggybacking including how it compares to walking unloaded and backpacking and how it relates to lumbar spine risk, through measurements of position, torques, and forces. Several variables were addressed, including maximal trunk flexion angle, external load torques, resultant muscle torque and force, joint reaction forces, vertical ground reaction forces, and gait variables. The primary purpose of the present study was to determine the maximal angle of trunk flexion across load conditions. A secondary purpose was to determine what torques were experienced and in which direction; these torques were not initially included in the variables but seemed increasingly important throughout the process. Other secondary problems included examining magnitude of the compressive, shear and resultant joint reaction forces at the lumbosacral joint.

Other Questions Regarding Piggybacking

Because piggybacking had never been investigated, there were some existing questions regarding using it as a training tool:

1) From walking with no load to carrying a person to carrying a backpack: what positions are occurring and what advice could be given for performing each, how does the position change and how will this affect torque and posture, or spine health?

2) Does piggybacking involve more or less risk of injury than backpacking an equal weight load?

Considerations

Prior to the pilot study, it was thought that piggybacking was more dangerous or risky than carrying a backpack of equal load. It was thought that the person being carried may feel further away from or as if they were falling off of the trunk of the carrier and that a backpack would fit snugly. If the trunk of the person being carried was indeed further away from the trunk of the carrier in comparison to a backpack, then their centre of gravity would be further from the axis of rotation and the moment arm of the line of gravity of the carried person would be longer. This would have resulted in their weight producing a greater amount of backward torque than the equally weighted backpack. As a result, the trunk would have exhibited more flexion during piggybacking to counteract the extensor torque. After the pilot study, this theory was considered incorrect and new hypotheses were considered.

Although the piggyback load was substantially heavier than the backpack load in the pilot study, the person being piggybacked was actually the preferred load to carry as vocalized by the participants. The trunks of the carriers were in a position of less flexion during the piggyback, even though the load was substantially greater. This may have been because part of the load was actually anterior to the axis of rotation; the arms and legs were wrapped around the carrier. The centre of gravity of the piggyback load was actually more anterior than the centre of gravity of the backpack load, which was concentrated totally posteriorly in the bottom third of the backpack. Also, the participants said that the piggyback passenger felt much more comfortable and almost lighter – definitely easier to carry although substantially heavier. Partly for these reasons, it was thought that the lumbosacral risk of piggybacking may be somewhat less than backpack carrying but also greater than carrying no load.

Null Hypotheses:

1) There are no differences between the maximum trunk flexion angles for each of the three load conditions.
2) There are no differences in the other kinematic, resultant load torque, internal lumbosacral joint torque and force, vertical ground reaction force, and gait variable values amongst the no load, piggybacking, and backpacking conditions.

$\mu_{NL} = \mu_{PB} = \mu_{BP}$

Alternative Hypotheses:

1) The trunk flexion angle exhibited during piggybacking is greater than this angle when carrying no load, and different than this angle during backpacking.
2) The other kinematic, resultant load torque, internal lumbosacral joint torque and force, vertical ground reaction force, and gait variable values are not equal amongst the no load, piggybacking, and backpacking conditions.

μ_{NL}, μ_{PB}, and μ_{BP} are not all equal

DELIMITATIONS

1) The participants were active males who resistance train regularly with heavy loads and were considered strong enough to experience only minimal risk during this study. The general population would likely be at greater risk if carrying the same magnitude of load due to having a less strength-trained musculoskeletal system.

2) A small sample size. However, this will be enough to provide the first basic biomechanical data and results regarding the biomechanics of the piggyback carry.

3) Model Assumptions:

a) The use of a static model. There were small accelerations as the trunk angle ranged approximately five degrees over each gait cycle, which resulted in an underestimation of the resultant muscle torque.

b) The sacral angle of 30 degrees did not alter while walking.

c) The resultant muscle line of force was parallel to the compressive joint reaction force and its moment arm was six centimeters.

d) Step 1 of the model identified the resultant torque; only the difference in trunk flexor and trunk extensor muscle force was able to be measured and therefore any co-contraction of the extensor muscles and flexor muscles that was present, increasing the compression at the lumbosacral joint, was unmeasurable and ignored. Therefore, the muscle force and calculated lumbosacral joint reaction forces were likely underestimated.

CHAPTER II

LITERATURE REVIEW

ANATOMY AND BIOMECHANICS OF THE BACK

THE VERTEBRAL COLUMN

The spine, or vertebral column, is composed of twenty four individual and two groups of fused vertebrae connected and stabilized by synovial facet joints, cartilaginous disc joints, and various ligaments and muscles (Alexander, 1985; Nordin & Frankel, 2012). Each pair of adjacent vertebrae and their connecting soft tissues are considered a motion segment and one functional unit of the spine (Hall, 2007; Nordin & Frankel, 2012). Together, these linked motion segments form the body's primary support structure, which links the upper and lower bodies while allowing motion in three dimensions and protecting the spinal cord (Hall, 2007). In its neutral position, the spine has four curved sections from superior to inferior as seen in Figure 2.1: the cervical, thoracic, lumbar, and sacrococcygeal curves (Hall, 2007; Moore et al., 2010). The cervical and lumbar curves are lordotic or concave anteriorly, and the thoracic and

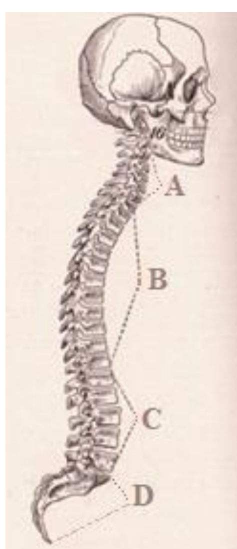


Figure 2.1: The four curves of the spine, alternating in lordotic and kyphotic curvature.
A) Cervical Spine B) Thoracic Spine C) Lumbar Spine D) Sacrococcygeal Spine
(Davison, 1908)

sacrococcygeal curves are kyphotic or convex anteriorly (Hall, 2007). The vertebrae in each section vary in characteristic structure and function (Hall, 2007). The lumbar region is of particular interest because of the current prevalence and medical costs related to low back pain in Canada today (Hall, 2007). This review will concentrate on the inferior lumbar spine, which bears the greatest loads and may be placed in a compromised position when a load is carried on the back.

These four curves increase the flexibility and weight-bearing capacity of the spine (Alexander, 1985; Hall, 2007). This flexibility helps the intervertebral discs absorb shock (Hall, 2007; Moore et al., 2010). These curvatures allow for optimal distribution of load throughout the spine, which enables higher loads to be supported without injury (Rousouly & Nnadi, 2010). When there is excess lumbar lordosis, the posterior elements of the spine are stressed by

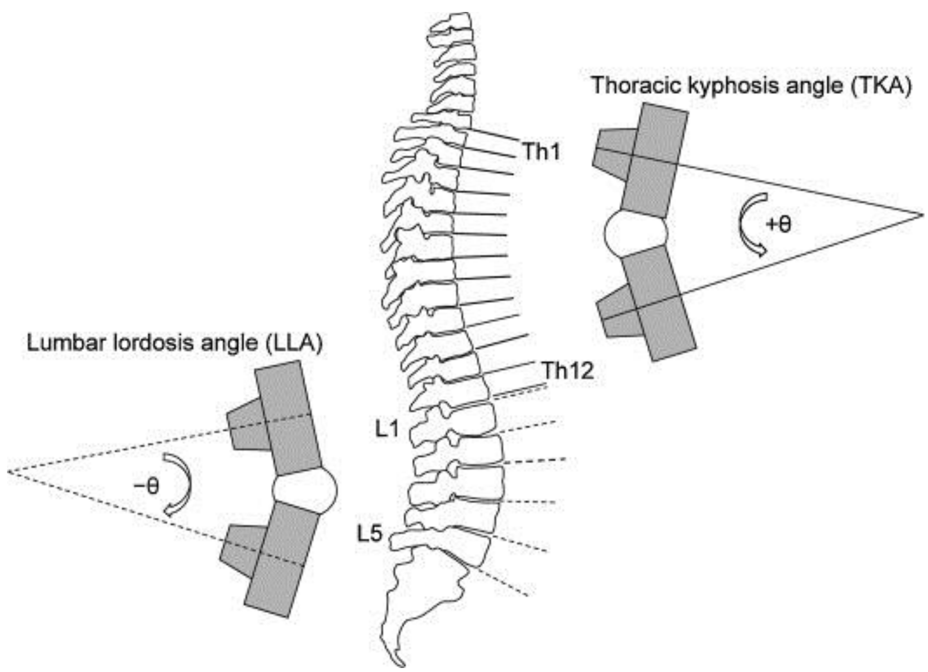


Figure 2.2: Illustration of the thoracic kyphosis and lumbar lordosis angles. The thoracic angle was measured between the superior surface of T1 and the inferior surface of T12 and the lumbar angle was measured between the superior surface of L1 and the inferior surface of L5. Reprinted from Archives of Gerontology and Geriatrics, 56/1, Miyazaki, Murata, Horie, Uematsu, Hortobagyi, & Suzuki, Lumbar lordosis angle (LLA) and leg strength predict walking ability in elderly males, 141-147, Copyright 2013, with permission from Elsevier.

compression while the anterior elements are placed on a stretch (Hall, 2007).

The amount of lumbar lordotic curve varies with the amount of thoracic kyphotic curve in order to allow proper upright posture; the angles of these curves can be seen in Figure 2.2 (Roussouly & Nnadi, 2010). The junction between these two curves is relatively flat with respect to the horizontal axis. Normal values for cervical lordosis, thoracic kyphosis and lumbar lordosis are approximately 40±9.7 degrees, 20 to 50 degrees, and 31 to 79 degrees respectively (Roussouly & Nnadi, 2010). Lumbar lordosis angle sometimes includes the lumbosacral joint, which is seen in Figure 2.3 and has a large angle of lordosis averaging 30 degrees (Roussouly & Nnadi, 2010). Across 49 healthy participants, the T1 to T12 surfaces angle of thoracic kyphosis was 45±8 degrees in males and 42±13 degrees in females, and the T12 to L5 surfaces angle of lumbar lordosis was 61±10 degrees in males and 58±10 degrees in females (Legaye, Duval-Beaupere, Hecquet & Marty, 1998).

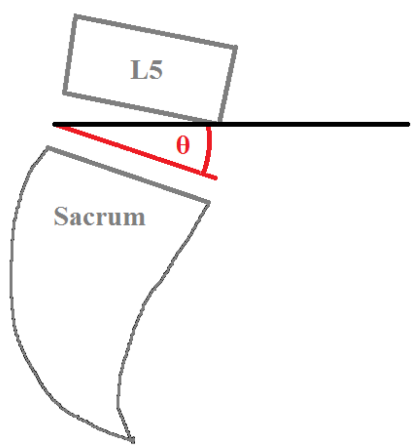


Figure 2.3: The lumbosacral angle. θ is equal to 30 degrees from the horizontal, on average.

The vertebrae can move in rotation or translation along each of the three primary axes; the mediolateral axis is the most relevant because the present study focuses on sagittal plane movements (Alexander, 1985; Nordin & Frankel, 2012). An axis of rotation is an imaginary line or pivot point about which rotation occurs in a two-dimensional plane perpendicular to the axis. Sagittal plane movement occurs either anterior-posteriorly, superior-inferiorly, or as rotation about the mediolateral axis which is oriented left to right. The primary variable measured in the present study will be trunk flexion, which is the forward rotation of the trunk around the mediolateral axis of rotation. In reality, there is no single axis of rotation, but a left-right axis through each of the intervertebral discs. The exact location of the lumbosacral joint axis alters slightly with joint position, but is typically found and will be assumed to be slightly posterior in the lumbosacral intervertebral disc (Bogduk, 1992; Nordin & Frankel, 2012). The location of this axis is illustrated in Figure 2.4 and will be vital to the lumbar spine force and torque calculations discussed later, as the length of each moment arm acting at the joint is dependent on the location of the axis.

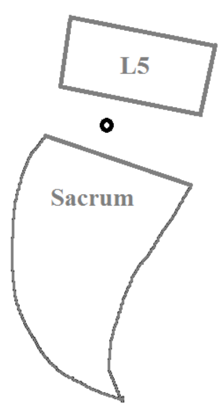


Figure 2.4: The lumbosacral mediolateral axis of rotation. It is slightly posterior in the centre of the intervertebral disc.

Lumbar Vertebrae

Each lumbar vertebra can be divided into two main parts: the vertebral body anteriorly and the vertebral arch posteriorly, as seen in Figure 2.5 below. The vertebral body bears most of the load that is supported by the spine, with greater loads supported by more inferior vertebrae

(Hall, 2007). Because of this, the vertebral bodies of lumbar vertebrae are thicker and broader than the bodies of cervical and thoracic vertebrae (Hall, 2007; Moore et al., 2010). The stress experienced by the vertebra is increased by muscle forces as well as any load carried by the upper body. Vertebral bodies have evolved for these demands and are strongest in compression (Nordin & Frankel, 2012). All of the forces on the vertebrae help determine the design of their inner structure in the form of bony pillars called trabeculae, which develop for strength and resistance along common lines of stress. Stress is equal to the amount of force experienced per unit of area. Therefore, the larger the area of the vertebral body supporting surface of a lumbar vertebra, the less stress the tissue will experience (Hall, 2007).

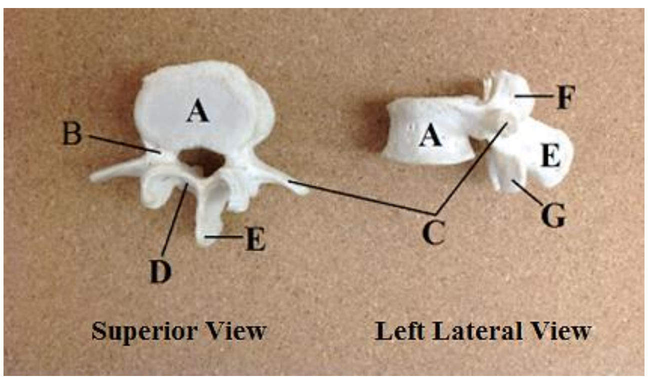


Figure 2.5: Superior V iew and Left Lateral View of a Lumbar Vertebra. A) Body B) Pedicle C) Transverse Process D) Lamina E) Spinous Process F) Superior Articular Process G) Inferior Articular Process

The vertebral arch protects the spinal cord and consists of the pedicles, laminae, and spinous process. The bilateral pedicles (B) connect the vertebral body (A) to the transverse processes (C). The bilateral laminae (D) bridge between the transverse processes and the spinous process (E), which is the most posterior bony prominence (Alexander, 1985; Moore et al., 2010). Lumbar vertebrae are recognizable by their large vertebral bodies and wide, paddle-like spinous processes. The vertebral arch and vertebral body surround the spinal canal to form the protective encasement for the spinal cord, and the space between the pedicles of successive vertebrae allows the horizontal passage of spinal nerve roots (Nordin & Frankel, 2012).

The bony projections from the neural arch serve several purposes. The length provided by the spinous process and transverse processes increases the moment arm and mechanical advantage of any attached muscles. The superior articular process and the inferior articular process are projections from the pedicle and lamina respectively and are important structures because they house the articulating facet surfaces (Moore et al., 2010). The narrow vertical structure connecting the superior articulating process (F) and the inferior articulating process (G) of the same vertebra is known as the pars interarticularis and is commonly fractured on either one or both sides of the body (Alexander, 1985; Hall, 2007). The size and angle of the bony processes vary with region of the spine, and determine the orientation of the facet joint surfaces, which limit the spinal movements possible in that region.

Joints and Mobility of the Spine

Every spinal movement occurs at joints of two different structure types. There are three joints between the two vertebrae of each motor segment: the right and left synovial facet joints between adjacent superior and inferior articular processes and the fibrocartilaginous disc joint between the two vertebral bodies (Hall, 2007; Moore et al., 2010), the latter of which will be discussed later. Each facet joint includes both hyaline cartilage facet surfaces, the synovial capsule, synovial fluid, and capsular ligaments (Alexander, 1985). Facet, or zygapophyseal, joints are important because they control and restrict movement, provide lateral stability of the motion segment, and may bear a significant amount of load depending on their orientation (Hall, 2007; Nordin & Frankel, 2012).

There is a great amount of loading of the facet joints when the spine is rotated and in flexion (Nordin & Frankel, 2012); the facet joints provide about 40 percent of the ability of the spine to resist torsion and shear (Hall, 2007). The facet joints resist up to 30 percent of a compressive load, with the most during hyperextension as the facet joints are directly compressed in this position (Hall, 2007; Levangie & Norkin, 2005). The facet surfaces that experience contact forces of the greatest magnitudes are at the lumbosacral joint (Hall, 2007). Most facet joints usually only bear weight temporarily as the spine changes positions, but the inferior articular process of the fifth lumbar vertebra bears load even when the spine is motionless and in the upright position (Moore et al., 2010), likely due to the large angle of the lumbosacral joint and greater shear component of body weight. It has been suggested that facet joint issues such as ligament sprains or articular surface damage are the root cause of chronic low back pain in many individuals (Hall, 2007).

The orientations of the articular process facet surfaces determine the direction of motion theoretically available to each vertebra, which varies with each curved section of the spine (Alexander, 1985; Hall, 2007; Moore et al., 2010; Nordin & Frankel, 2012). The surfaces of the cervical facet joints are oriented relatively horizontally and obliquely (Moore et al., 2010), which allows flexion-extension, lateral flexion (Hall, 2007; Moore et al., 2010; Nordin & Frankel, 2012), and rotation (Hall, 2007; Nordin & Frankel, 2012). The surfaces of the thoracic facet joints are between parallel to and 20 degrees to the sagittal plane and 60 degrees to the horizontal plane (Nordin & Frankel, 2012); the thoracic spine is most mobile in lateral flexion, rotation, and some flexion-extension. The surfaces of the lumbar vertebra facets are vertical and at 45 degrees to the frontal plane, which allows flexion-extension, lateral flexion, and almost no rotation (Nordin & Frankel, 2012). There is also a large amount of variation in the orientation of the facet joints, and therefore range of motion, within and between individuals (Nordin & Frankel, 2012). Every movement of the trunk is experienced by more than one motion segment (Hall, 2007; Nordin & Frankel, 2012). Due to the interlocking structure of the vertebrae, the spine experiences slight lateral flexion to the same side when it experiences rotation (Hall, 2007).

The upper lumbar spine exhibits a greater range of motion in flexion and extension in comparison to the lower lumbar spine, which exhibits a greater range of motion in lateral flexion. However, the difference is slight for lateral flexion; each lumbar motion segment can exhibit approximately five to six degrees of lateral flexion (Alexander, 1985; Hall, 2007; Nordin & Frankel, 2012). Many sports skills require frequent lumbar hyperextension, which can result in the spine being able to exhibit twenty times the normal angle of the curve (Hall, 2007). The range of motion of rotation is limited to approximately two to three degrees between successive lumbar vertebrae (Alexander, 1985; Hall, 2007), decreasing from nine degrees in the upper thoracic to two degrees in the lower lumbar spine due to interlocking articular processes (Nordin & Frankel, 2012). Because the facet joints limit rotation in this region, excessive lateral compression forces on the facet joints are transmitted to the neural arch if the lumbar vertebrae rotate more than approximately three degrees with respect to each other (Alexander, 1985).

As a whole spine, the normal flexion-extension range of motion in percentage is 4% for sitting, 11% for walking normally or down stairs, 13% for walking upstairs, 39% for moving from sitting to standing, 52% for squatting, and 59% for bending (Bible et al., 2010). Of the tasks that have been quantified, the greatest range of motion in the lumbar spine was needed for picking up an item from floor level (Hall, 2007).

The Lumbosacral Joint

At the inferior aspect of the spine, the fifth lumbar disc articulates with sacrum, which is known as the L5-S1 joint, lumbosacral joint, or lumbosacral junction (Alexander, 1985; Moore et

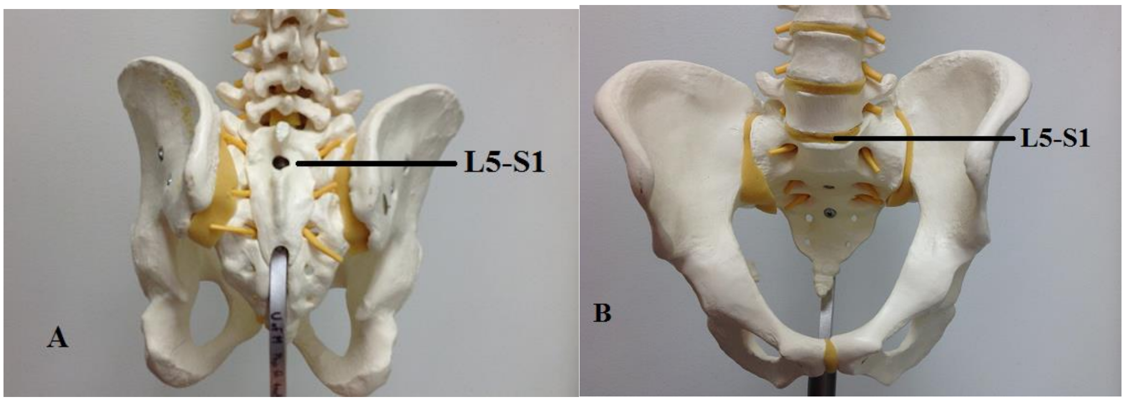


Figure 2.6: The Lumbosacral Joint. A) posterior view B) anterior view

al., 2010). The lumbosacral joint is seen in Figure 2.6, and is the inferior endpoint of the lumbar lordotic curvature (Moore et al., 2010). The fifth lumbar vertebra is wedge shaped, taller anteriorly than posteriorly, so as to form the inferior portion of the lordotic curve and the transition from vertebral column to sacrum (Levangie & Norkin, 2005). The lumbosacral facet surfaces are oriented more obliquely than those of most lumbar vertebrae, which allow for approximately three degrees of lateral flexion and a greater range of motion of rotation than other joints in the lumbar region: up to five degrees (Hall, 2007; Nordin & Frankel, 2012). The lumbosacral range of motion is much greater for flexion-extension, as it reaches up to approximately twenty degrees (Hall, 2007; Nordin & Frankel, 2012).

The angle between the superior surface of the sacrum and the horizontal axis is considered the sacral slope or lumbosacral angle and is illustrated in Figure 2.7. The average angle of the lumbosacral junction is approximately 30 degrees to the horizontal during relaxed standing, but ranges due to individual anatomy and posture (Levangie & Norkin, 2005; Nordin & Frankel, 2012). Russouly & Nnadi (2010) stated that the sacral angle normally averages 40 degrees but ranges from 20 to 65 degrees. An increase in the size of this angle affects the lumbar curve by increasing its lordosis, as well as the amount of shear force at the lumbosacral joint as illustrated in Figure 2.8 (Benedek & Villars, 1974; Levangie & Norkin, 2005).

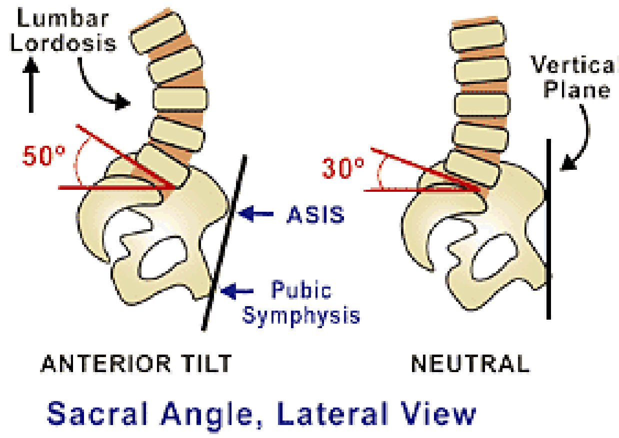


Figure 2.7: Normal Sacral Angle and Anterior Pelvic Tilt. The pelvic tilt is measured by the relative anterior-posterior location of the anterior superior iliac spine and the pubic symphysis. Reproduced with permission from Dalton, E. (2014).

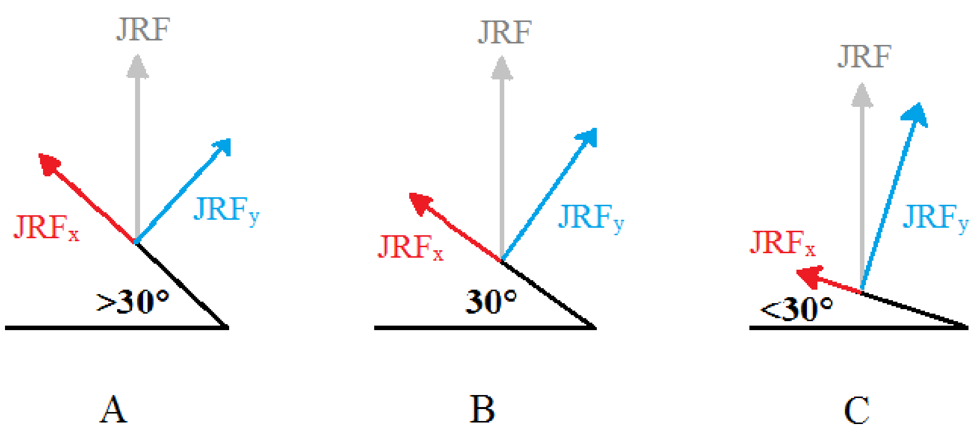


Figure 2.8: The magnitude of the shear component of the lumbosacral joint reaction force depends on the sacral angle. A) Sacral angle greater than 30°: a relatively greater shear component and lesser compressive component, B) Average sacral angle at 30° to the horizontal, C) Sacral angle less than 30°: a relatively lesser shear component and greater compression component.

The greater the lumbosacral angle, the less the proportion of compressive force and the greater the proportion of the shear force within the contact force. Trunk hyperextension and lumbar hyperlordosis can increase the angle of anterior tilt and the magnitude of anterior shear forces (Alexander, Adv Bio, 2010). The pelvis is not known to tilt anteriorly with trunk flexion until after the first 20 to 30 degrees of trunk flexion (Chaffin, Anderson, & Martin, 1999). Therefore, pelvic tilt would not be expected to alter with the relatively minimal trunk flexion that is caused by carrying a moderate load; the pelvis would likely remain at an angle close to 30 degrees with respect to the horizontal axis while walking with a load. Anterior pelvic tilt and posterior pelvic tilt are illustrated in Figure 2.9.

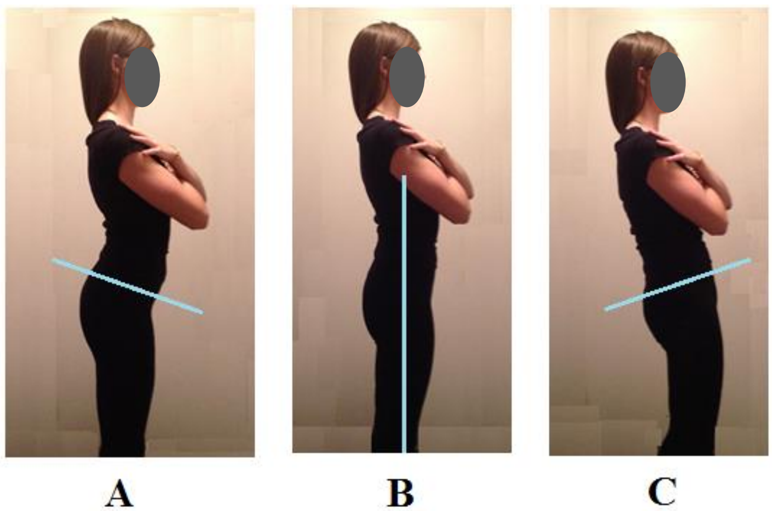


Figure 2.9: Examples of pelvic tilt: A) anterior pelvic tilt, B) normal pelvic posture, C) posterior pelvic tilt.

Intervertebral Discs

There is another joint between successive vertebrae and within the lumbosacral joint but unlike the synovial facet joints, this joint is fibrocartilaginous. Each pair of vertebral bodies is separated by a fibrocartilaginous disc, and collectively, the wedge shapes of the discs allow for the spinal curvatures (Hall, 2007). Each disc is composed of an outer fibrocartilage annulus fibrosus and an inner gel-like nucleus pulposus, as illustrated in Figure 2.10 (Alexander, 1985; Hall, 2007; Moore et al., 2010).

The annulus fibrosus is able to withstand large amounts of bending and torsion because of its crisscross collagen fibre arrangement that provides strength against tension in many directions (Levangie & Norkin, 2005; Nordin & Frankel, 2012). There are approximately 90 (Hall, 2007) concentric bands of collagen angled at 120 degrees to each other in ten to twelve to twenty alternating layers called lamellae (Bogduk &Twomey, 1987; Bogduk,1991). Because of

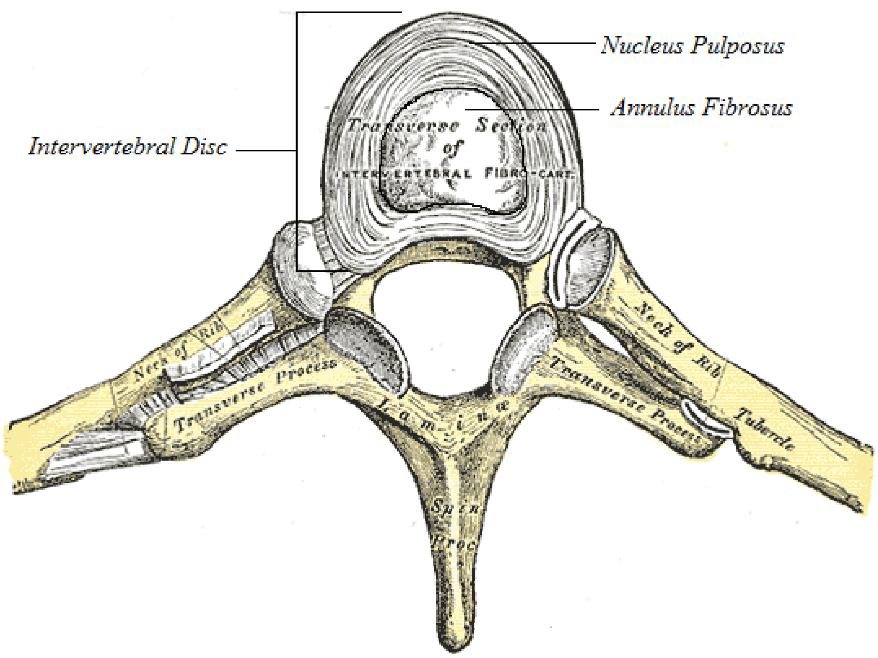


Figure 2.10: Structure of the intervertebral disc- the inner nucleus pulposus and outer annulus fibrosus. (Gray`s Anatomy, 1918)

this arrangement, the annulus fibrosus is mechanically designed to better resist rotation than compression, tension, or shear (Benedek &Villars, 1974; Hall, 2007). The bonding of the collagen fibres and layers are crucial to the function of the disc. During loading and degeneration, the organization and orientation of these fibres and layers alter. The bending stress produced by flexion on the annulus fibres is 450 times greater than the twisting stress of a rotation of equal angle (Hall, 2007).

Immediately superior and inferior to the disc are cartilaginous end plates, which are the attachment points for the elastic fibres of the annulus fibrosus and are themselves very weakly attached to the surfaces of the vertebral bodies (Alexander, 1985; Levangie & Norkin, 2005). At the centre of the annulus fibrosus is the nucleus pulposus. The nucleus pulposus of the lumbar disc is slightly posterior in comparison to its centered position in the cervical and thoracic vertebral discs (Nordin & Frankel, 2012). The nucleus pulposus is primarily composed of water, with some collagen and proteoglycans, which aid in retaining the water component.

Loading the disc not only alters the arrangement of the collagen fibres of the annulus fibrosus, but also changes the internal environment of the disc. For example, the intrinsic pressure and proportion of water will alter. There is approximately $10N/cm^2$ of intrinsic pressure in the unloaded intervertebral disc (Nordin & Frankel, 2012) and the vertebral discs usually make up one quarter of the height of the spine (Hall, 2007). When a load is supported by this non-compressible gel, the gel is resisted by the vertebral endplates above and below it, and the annulus fibrosus on every side. In compression, the intrinsic disc pressure is approximately 150 percent of the external load per unit area (Nordin & Frankel, 2012). This compression makes the disc bulge laterally and the annulus experiences a circumferential tensile stress of approximately 400 to 500 percent of the applied longitudinal compressive load (Nordin & Frankel, 2012). The disc loses water content until the internal pressure is equal to the external pressure. As this happens, the discs decrease in height and more of the load is supported by the facet joints (Hall, 2007). Over the course of one day, the spine will decrease in height by approximately two centimeters due to water loss; more than one centimeter is lost in the first half an hour of erect posture (Hall, 2007). Once the load is relieved, water quickly reabsorbs into healthy discs.

Because the discs have greater volume and height in the morning, the spine is stiffer and there is a higher risk of injury (Hall, 2007).

The discs rely on the pumping action caused by water moving during changes in position for nutrient and waste transport because they do not have a blood supply. The lumbosacral intervertebral disc is wedge shaped, with a large superior surface area and a small inferior surface area (Levangie & Norkin, 2005). The height of the disc decreases due to compression over time, especially posteriorly, which may lead to crushing of the synovial capsule and pain (Benedek & Villar, 1974). Degeneration decreases the shock absorbing and water absorbing capacities of the disc, and is most common at the lumbosacral disc because its position causes it to endure the greatest mechanical stresses (Hall, 2007).

The intervertebral discs collectively aid in restraining excessive motion, bearing load and distributing load (Nordin & Frankel, 2012). The discs support flexion, extension, lateral flexion, and rotation about the three primary axes (Alexander, 1985). The discs allow the adjacent superior vertebra to rotate in the sagittal plane over their supporting nucleus centre while the facet joint guides the movement (Hall, 2007). During bending, whether it is flexion, extension or lateral flexion, one side of the disc experiences compressive stress while the other side of the disc experiences tensile stress. Therefore, when the spine is flexed, the anterior aspect of the intervertebral disc experiences increased compressive forces and the posterior aspect experiences increased tensile forces; the nucleus pulposus is simultaneously forced posteriorly into the tensed annulus tissue. When the spine is loaded, these forces within the disc would theoretically increase because of both the added load and the increased likelihood of trunk flexion.

The discs resist 40 percent of rotational torsion and shear stress, equal to the resistance provided by the facet joints, with some of the shear often caused by the torsion itself (Hall, 2007). During axial compression, the intervertebral discs support approximately 80% of the load while the facet joints share 20 to 30 percent of the load; these proportions vary with altered mechanics (Hall, 2007; Levangie & Norkin, 2005). They become less successful in absorbing shock with injury and age (Hall, 2007).

Ligaments of the Spine

The lumbar spine receives a significant amount of support and stability against excessive movements by the forces created by ligamentous structures, most seen in Figure 2.11. These structures provide passive resistance against any movement that causes the ligament to experience tension. The ligaments that support the lumbar spine include, from anterior to posterior, the anterior longitudinal ligament, posterior longitudinal ligament, ligamentum flavum, intertransverse ligament, capsular ligaments, interspinous ligaments, and supraspinous ligaments (Alexander, 1985; Hall, 2007).

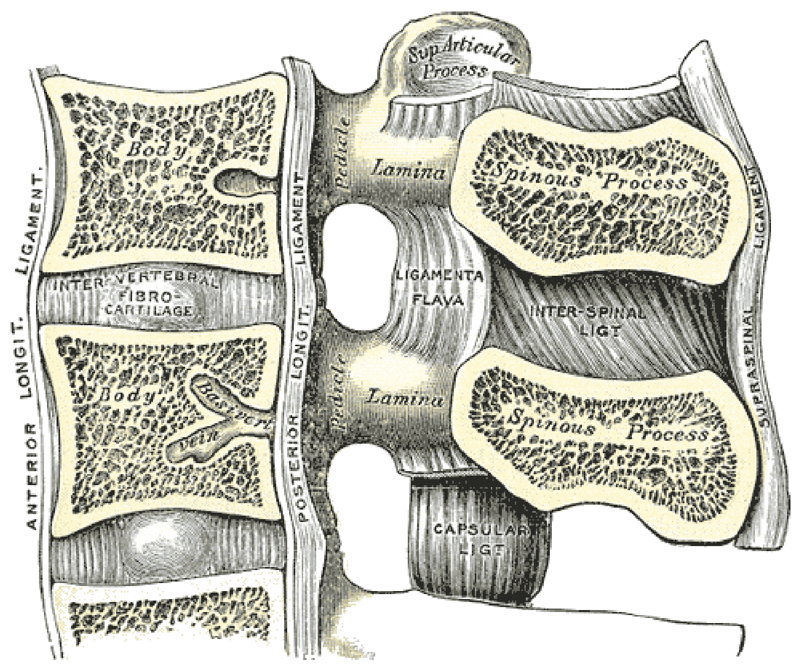


Figure 2.11: Ligaments that support the lumbar spine. (Gray`s Anatomy, 1918)

The anterior longitudinal ligament and posterior longitudinal ligament each run vertically along the antero-lateral and posterior surfaces of the vertebral bodies (Hall, 2007), respectively, from the body of the axis to the superior sacrum (Levangie & Norkin, 2005). The anterior longitudinal ligament is considered several times as strong as the posterior longitudinal ligament, although they both have deep and superficial fibres and a similar moment arm (Hall, 2007; Levangie & Norkin, 2005). The deep fibres of the anterior longitudinal ligament blend with the fibres of the annulus fibrosus to provide extra support for the anterior disc and anterior intervertebral joint (Levangie & Norkin, 2005). The lumbar portion of the anterior longitudinal ligament has demonstrated up to 676 N of axial tensile strength. The lumbar posterior longitudinal ligament specifically is known not to be a strong spine stabilizer due to its narrow width and short moment arm; its lumbar portion can only produce approximately 300 N of axial tension strength (Bogduk, 1989; Levangie & Norkin, 2005). Although some of its fibres attach to the borders of the vertebral endplates and annulus fibrosus, it lacks in providing support for the intervertebral discs, leading to injuries in the posterior disc (Levangie & Norkin, 2005).

Ligamentum flavum is located along the posterior aspect of the vertebral canal and spans longitudinally between each pair of adjacent vertebral arches, connecting the lamina of adjacent vertebra (Hall, 2007; Nordin & Frankel, 2012). This ligament is particularly well-developed in the lumbar region (Levangie & Norkin, 2005). The ligamentum flavum has high elastin content in comparison to the very high collagen fibre concentration of most other ligaments, which allows for its unique function (Hall, 2007; Nordin & Frankel, 2012). Because collagen fibres limit the extensibility of ligaments and elastin fibres provide extensibility, its high elastin concentration allows ligamentum flavum to be under constant tension when upright; it consistently exerts a compressive force on each intervertebral disc at a distance from the axis of rotation.   This constant moment causes pre-stress on the disc, which increases the weight-bearing capacity of the spine by contributing to intradiscal pressure (Hall, 2007; Nordin & Frankel, 2012).

The intertransverse ligament connects the transverse processes, and transitions into broad, membrane-like sheets in the lumbar region (Hall, 2007; Levangie & Norkin, 2005). One source discussed the blending of this ligament into the iliolumbar ligament inferior to the third lumbar vertebra (Levangie & Norkin, 2005). Three strong bands of the iliolumbar ligament extend from the transverse processes of the fourth and fifth lumbar vertebrae to the ilium of the hip bone. They aid in stabilization of the fifth lumbar vertebra, preventing anterior displacement of the vertebra as well as tension against excessive flexion, extension, axial rotation, and lateral bending at the lumbosacral joint (Levangie & Norkin, 2005).

The capsular ligaments make up the facet joint capsules, spanning from the superior articular process of one vertebra to the inferior articular process of the vertebra above it. The capsular ligaments are well-developed at the thoracolumbar region as the curvature changes (Levangie & Norkin, 2005). These ligaments aid in providing spinal stability and resistance to motion. Just like the facet joints that they encapsulate, the movements of the spine that are resisted by the capsular ligaments vary by region of the spine. In the lumbar region, they work with the fibres of the annulus fibrosus to resist axial rotation. They are most vulnerable to hyperflexion, possibly providing more resistance than any of the posterior ligaments and failing after the supraspinous and interspinous ligaments, which are usually the first to fail in high load hyperflexion (Levangie & Norkin, 2005).

The lumbar interspinous ligament is composed of broad, thick, sheet-like structures that span between the entire antero-posterior lengths of the spinous processes of two adjacent vertebrae (Hall, 2007; Levangie & Norkin, 2005). Its fibre orientation has been debated, from parallel to oblique. Other than to resist flexion, the additional functions of the interspinous ligament have been suggested as to resist end-range extension, resist posterior shear of a vertebra on the vertebra below it, or unfortunately to produce anterior shear during full flexion (Levangie & Norkin, 2005). The interspinous ligament is not as strong as the anterior longitudinal ligament, posterior longitudinal ligament, or ligamentum flavum in axial tension, although its moment arm is actually greater. It may be able to provide up to 200 N of tensile strength (Bogduk, 1989). This ligament aids in stability of the lumbar spine, and is known to degenerate with age (Levangie & Norkin, 2005).

The supraspinous ligament is a fibrous cord-like structure that attaches to the posterior aspect of each spinous process, spanning from the seventh cervical vertebrae to either L4, L5, or the sacrum (Hall, 2007; Levangie & Norkin, 2005). Its fibres merge with the fibres of the lumbodorsal fascia and muscle insertion points as it reaches the lumbar region. This ligament resists the separation of the spinous processes as the trunk moves into flexion (Levangie & Norkin, 2005). The supraspinous ligament has been speculated to house mechanoreceptors specialized to lead to the recruitment of muscles that stabilize the spine, like the multifidus (Levangie & Norkin, 2005).

The ligaments that exert the greatest amount of force to support joint structures vary with the position of the spine, whether it is flexion, rotation, extension, or lateral flexion. Trunk flexion, which can be induced by load carriage (Birrell et al., 2007; Gillet el al., 2006) causes a forward moment about the mediolateral axis of rotation, located within the intervertebral disc. If the body reaches extreme flexion, stress is placed on the ligamentous structures posterior to the axis of rotation, which can produce a backward moment when tensed. The ligaments that experience the greatest stress during trunk flexion are the supraspinous and interspinous ligaments, then the capsular ligaments and ligamentum flavum (Levangie & Norkin, 2005; Nordin & Frankel, 2012). The interspinous and supraspinous ligaments are the most stressed during extreme flexion due to their longer distance from the axis of rotation, and are therefore usually the first to fail (Levangie & Norkin, 2005). While intact, the interspinous and supraspinous ligaments together strongly resist anterior shear during spinal flexion (Alexander, 1985). The posterior longitudinal ligament and supraspinous ligaments are also posterior to the axis and are therefore also stressed during trunk flexion (Hall, 2005; Levangie & Norkin, 2005).

Each ligament is most stressed when its attachment points are separated. Rotation causes the greatest stress to the capsular ligaments of the facet joints (Nordin & Frankel, 2012). The spine may experience loaded rotation when trying to control turning movements or the loading or unloading of a backpack or person. The anterior longitudinal ligament experiences the greatest tensile force when the spine is in extension, and the contralateral ligamentum flavum and capsular ligaments experience the greatest tensile forces during lateral flexion (Nordin & Frankel, 2012). Due to its lateral position, the intertransverse ligament on one side is also stretched, while the other is compressed, during lateral flexion (Levangie & Norkin, 2005).

Stability of the Spine

There are three subsystems that contribute to the stability of the spine: passive structures, active structures and neural control. Passive stability is provided by several structures that have been discussed above, including bones, intervertebral discs, ligaments and joint capsules, as well as the passive properties of muscles and tendons (Kavcic, Grenier, & Stuart, 2004). The passive system is only able to provide control for a load equivalent to the weight of the head if it is not supplemented. This assistance is derived partly from the second subsystem, active stability provided by the muscles that support the posture of the spine (Kavcic et al., 2004). These muscles will be discussed in the following section. The nervous system is responsible for acknowledging when stability needs to be increased via receptors surrounding the spine, and coordinating the appropriate response using motor neurons to activate the appropriate muscles (Kavcic et al., 2004). These three subsystems work together to maintain stability of the spine; a deficit in any aspect can lead to decreased stability and increased risk of injury.

Muscles

Many muscles work together to control the movement, balance and posture of the spine, and to maintain its curvatures (Moore et al., 2010). These muscles can be divided into anterior trunk muscles that contract to produce trunk flexion, posterior trunk muscles that contract to produce trunk extension, and lateral trunk muscles. The trunk flexors and extensors coordinate ipsilateral contraction to produce trunk lateral flexion and two of the trunk flexors, the internal and external obliques, coordinate contralateral contractions to create trunk rotation (Moore et al., 2010). The trunk muscles coordinate contraction to increase spinal stability and intradiscal pressure, with the assistance of ligaments such as the ligamentum flavum (Nordin & Frankel, 2012; Wilke, Wold, Claes, Arand, & Wiesend, 1996).

The primary trunk flexor muscles are illustrated in Figure 2.12 and are the rectus abdominis, external abdominal obliques, internal abdominal obliques, and transversus abdominis from most superficial to deep (Alexander, 1985; Hall, 2007; Nordin & Frankel, 2012). These muscles are important because they are able to co-contract with the trunk extensors to simply stabilize the trunk or can exert a forward moment about the spine to maintain posture if the trunk has a tendency to rotate backwards. They also aid in resisting excess lumbar lordosis (Moore et al., 2010). Many researchers also previously believed that the abdominal muscles helped to

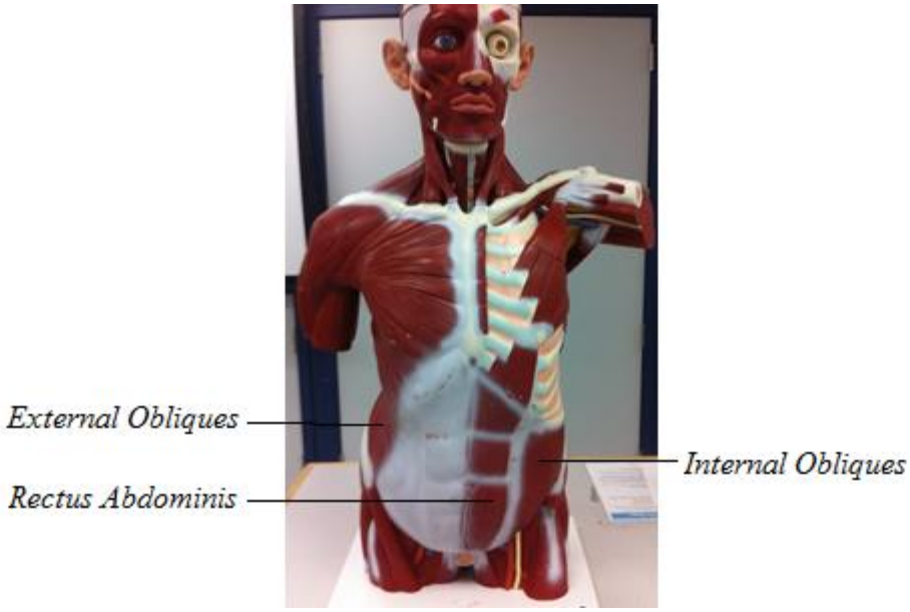


Figure 2.12: The abdominal muscles – the rectus abdominis, external oblique, and internal oblique. The left side of the model displays deeper structures. Transversus abdominis is below the surface and therefore not visible.

support load, which was known as the abdominal balloon theory (Hall, 2007; Levangie & Norkin, 2005).

The abdominal balloon theory suggested that the trunk flexors assisted the vertebral column by creating a pressurized abdominal cavity that supported a portion of the weight of the head, arms, and trunk (Hall, 2007; Levangie & Norkin, 2005; McGill & Norman, 1986). The weight of the upper body would have then been borne by both the supporting balloon in the anterior trunk and the vertebral column in the posterior trunk. However, this theory is now thought to be invalid because contraction of the abdominal muscles decreases the distance between their origins and insertions and would therefore add to the flexor moment supported by the spine. For example, the rectus abdominis would pull the ribs closer to the pubis, which would create compressive forces on the vertebral column. In fact, when intraabdominal pressure is increased, there is a proportionate increase in trunk extensor moment and an increase in pressure in the lumbar discs. In conclusion, the abdominal balloon theory is not supported by recent authors such as Chaffin et al. (1999), Hall (2007), Levangie and Norkin (2005), and McGill et al. (2013).

Instead, it is important to be aware that when a heavy or unstable load is carried, a flexor torque produced by the abdominals increases intra-abdominal pressure but also increases the flexor torque, and is offset by a portion of the extensor torque. Although the loads carried in this study will not be excessive, the lumbosacral resultant torque estimation may be underestimated because of offsetting due to co-contraction of the trunk flexors and extensors.

Isometrically contracting the muscles of the torso increases stability but also increases the load on the spine because many of the muscles generate compressive forces on the spine (McGill et al., 2009a). During a cable workout with a mean load of 5.4±1.6 kg, consciously bracing the torso increased the L4-L5 compression force from 3345 N to 4185 N, a difference of 840 N (McGill et al., 2009b). Torso bracing increased the shear forces by 32 to 55 percent in moderate resistance training exercises, while it increased the muscle compression forces from 29 to 46 percent (McGill et al., 2009a).

The muscles of the inferior trunk are associated with a thickened tissue known as either the thoracolumbar or lumbodorsal fascia. This fascial structure consists of three layers as seen in Figure 2.13, with the posterior and middle encasing the erector spinae and multifidi extensor muscles and the anterior layer derived from the fascia of the quadratus lumborum (Levangie & Norkin, 2005). The quadratus lumborum muscle is located in the lateral trunk region; it performs lateral flexion of the spine unilaterally (Levangie & Norkin, 2005), extension of the spine bilaterally (Tortora & Nielsen, 2009), and is an important stabilizer in the frontal and horizontal planes (Levangie & Norkin, 2005). The thoracolumbar fascia is a dominant structure in the dorsal region and has attachment points at many structures, such as: thoracic, lumbar and sacral spinous processes and supraspinous ligaments, the posterior superior iliac spine, iliac crest, posterior longitudinal ligament, lumbar transverse processes and lumbar intertransverse

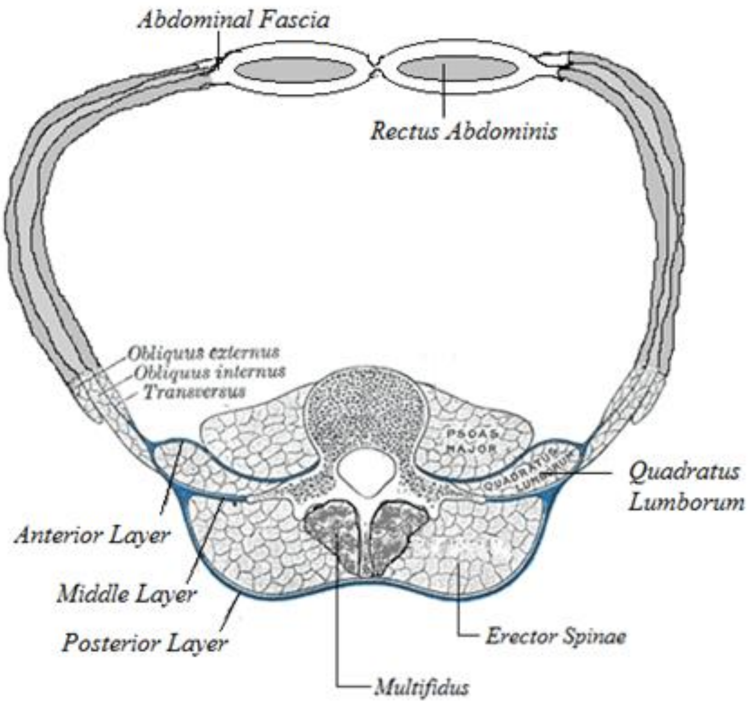


Figure 2.13: The layers of the thoracolumbar fascia. (Gray`s Anatomy, 1923)

ligaments (Levangie & Norkin, 2005). Its fibres blend with the latissimus dorsi muscle superiorly and the contralateral gluteus maximus inferiorly. Because it forms the lateral raphe at the lateral portion of the erector spinae, which gives rise to the internal abdominal oblique and transversus abdominis muscles, the lumbodorsal fascia can provide stability via intra-abdominal pressure (Levangie & Norkin, 2005). This structure as a whole transmits tension to and from its attachments and may help the trunk extensors to resist load. When muscles that attach to the lumbodorsal fascia are tensed, they transmit tension and “provide a pathway for the mechanical transmission of forces between the pelvis and the trunk” (Levangie & Norkin, 2005, p. 170).

Muscles that Extend the Trunk at the Lumbosacral Joint

The primary trunk extensor muscles vary slightly by region of the spine. The most commonly cited extensors of the lumbar spine are the erector spinae group (Alexander, 1985; Bogduk et al., 1992; Hall, 2007; Moore et al., 2010; Nordin & Frankel, 2012) and multifidi (Bogduk et al., 1992; Hall, 2007; Moore et al., 2010; Nordin & Frankel, 2012). Other muscles such as rotatores, interspinalis (Hall, 2007), and intertransversarii (Hall, 2007; Nordin & Frankel, 2012) may also be cited as lumbar extensors. Bogduk et al. created a lumbar spine model as anatomically accurate as possible, based on radiographs and finely detailed cadaver dissections, while considering the results of previous research (1992). They represented the lumbar extensors via 29 erector spinae and 20 multifidus muscle fascicles, some attaching within the lumbar spine and some attaching both superior and inferior to the lumbar spine, which all resulted in lumbar spine extension upon contraction.

The erector spinae muscle group is composed of the iliocostalis, longissimus, and spinalis muscle columns from lateral to medial as seen in Figure 2.14. Portions of these columns are named for their location, described as capitis, cervicis, thoracis/dorsi, or lumborum corresponding to the skull, cervical, thoracic, and lumbar regions. All of these three muscle columns originate at the posterior iliac crest, posterior sacrum, lower lumbar spinous processes, sacroiliac ligaments, and the supraspinous ligament via a broad tendon (Moore et al., 2010; Bogduk et al., 1992). However, each column has a different insertion superiorly. Iliocostalis inserts on the lower rib angles and transverse processes of cervical vertebrae, and has all but a capitis region. Longissimus inserts in the cervical and thoracic regions, between the tubercles and angles of ribs and transverse processes of vertebrae, and to the mastoid process of the skull (Moore et al., 2010). Spinalis inserts on the spinous processes of the upper thoracic vertebrae and to the cranium (Moore et al., 2010). Both longissimus and spinalis only have fascicles in the dorsi, cervicis, and capitis regions, but because they originate inferior to the lumbosacral joint, they also provide extension moments at that joint and are essential to the lumbar spine model. The erector spinae aponeurosis transmits 40 to 80 percent, depending on the level of the lumbar spine, of the total extensor moment acting on the lumbar spine from fibres that do not arise until the thoracic region (Bogduk et al., 1992).

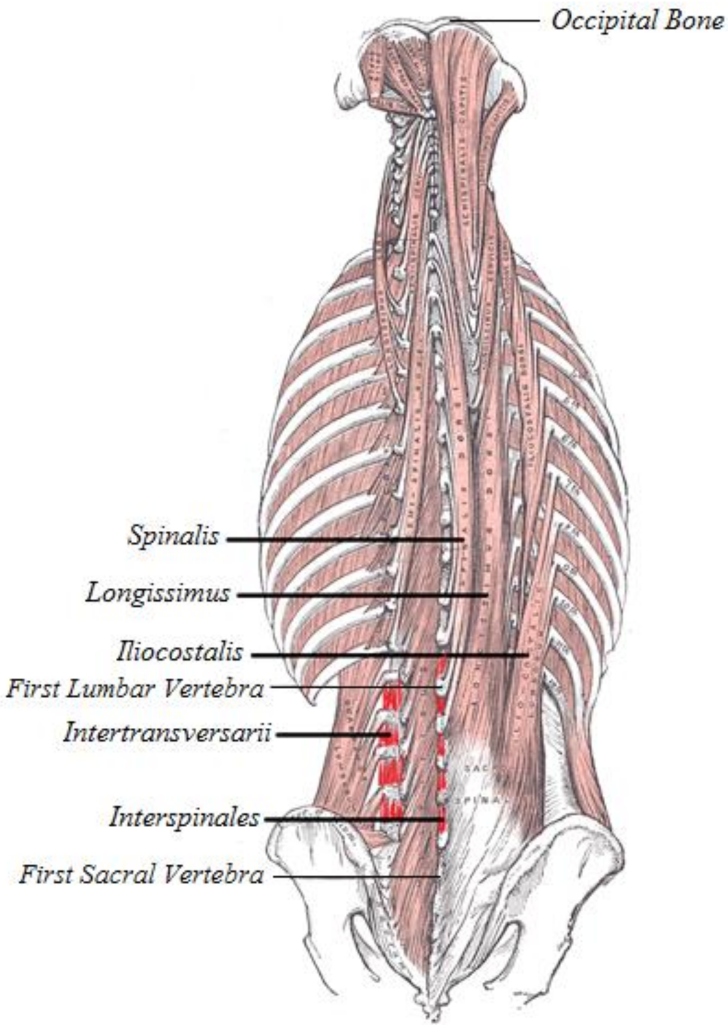


Figure 2.14: The Trunk Extensor Muscles. Iliocostalis, longissimus, spinalis, intertransversarii and interspinales. (Gray`s Anatomy, 1918)

The lumbar erector spinae are better discussed as functional groupings. The superficial erector spinae are essential to active trunk extension while the deep erector spinae play an important role in actively resisting anterior shear. The superficial layer consists of the portions of the longissimus thoracis and iliocostalis lumborum that have muscle bellies in the thoracic region, which more specifically pass upwards from the lower lumbar spinous processes, sacrum, and iliac crest to the transverse processes of the thoracic vertebrae and ribs (Levangie & Norkin, 2005). Their muscle bellies are located in the lateral thoracic region and they have long tendons in the lumbar region, resulting in a comparatively long moment arm and almost ideal line of pull to bilaterally produce extension of the thoracic and lumbar spines. These muscles eccentrically contract to control trunk flexion against gravity, assisted by the lumbodorsal fascia and the posterior ligaments (Levangie & Norkin, 2005). Levangie and Norkin (2005) suggested that once the body moves past two thirds of the range of motion of maximum flexion, the erector spinae only contribute passively via their elastic fibres and that the posterior ligaments and lumbosdorsal fascia take over the resistance to gravity. They also suggested that the supraspinous and interspinous ligaments actually have longer moment arms than the erector spinae, and therefore have a greater mechanical advantage (Levangie & Norkin, 2005).

The deep layer of the erector spinae is made up of many individual fascicles with common tendinous insertions, and may be considered separate from the superficial layer. The deep layer consists of the deeper and shorter portions of the longissimus thoracis and iliocostalis lumborum (Levangie & Norkin, 2005). The fibres run anteriorly and medially from the iliac crest as they move superiorly to the transverse processes of the lumbar vertebrae. Unfortunately, their moment arm is very small. However, although the muscles of this layer are not effective spinal extensors, they are very effective producers of dynamic posterior shear forces. They provide dynamic posterior shear to resist the anterior shear force caused by the sacral angle, body weight, and ground reaction forces. Once the spine is flexed, however, the line of pull of these muscles is not sufficient to effectively perform this very important protective function (Levangie & Norkin, 2005).

The transversospinales muscle group includes the semispinalis, multifidus, and rotatores muscles. These muscles originate at the transverse processes of each vertebra and insert at the spinous process, lamina and ligaments that they pass en route to the spinous process of a more superior vertebra (Moore et al., 2010). Semispinalis does not act on the lumbar spine, but rotatores and multifidus stabilize the lumbar spine and perform local extension bilaterally or rotation of the spine unilaterally (Moore et al., 2010). Multifidus is illustrated in Figure 2.15. Some researchers have suggested that rotatores has such a small moment arm and size that it performs more of a proprioceptive role (Levangie & Norkin, 2005). In comparison to the thoracic and cervical multifidus muscles, the lumbar multifidus is better designed for spinal extension because it has a greater cross-sectional area and its line of pull is more vertical and less oblique (Levangie & Norkin, 2005); the main purpose of multifidus is extension of the spine (Macintosh & Bogduk, 1986). When producing extension, the lumbar multifidus increases lumbar lordosis, which increases compressive forces on the posterior intervertebral discs due to the position of the spine (Levangie & Norkin, 2005). Contraction of the multifidus is coordinated to resist the trunk flexion moment created by the abdominals (Macintosh & Bogduk, 1986). The multifidus muscle is a very important spinal extensor, creating approximately 20 per cent of the extensor torque at the lumbosacral level (Bogduk et al., 1992).

The intertransversarii muscles, seen in Figure 2.14, are not always included in lumbar spine models but are potentially capable of producing extension of the lumbar spine. They connect adjacent transverse processes in cervical and lumbar vertebrae (Moore et al., 2010), and are therefore posterior to the flexion-extension axis of rotation. Moore et al. (2010) stated that

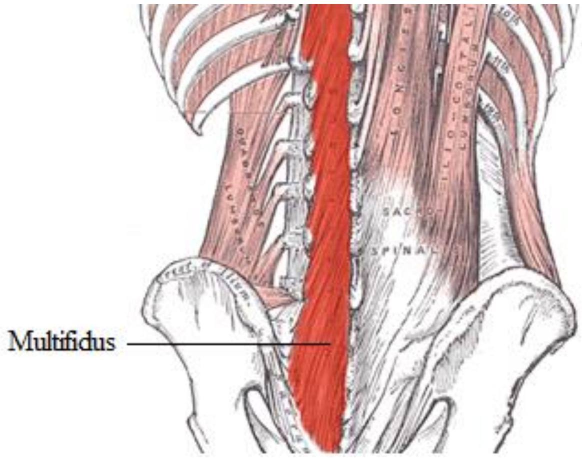


Figure 2.15: Multifidus. The multifidus muscle extends or rotates the trunk upon bilateral or unilateral contraction, respectively. (Gray`s Anatomy, 1918)

the intertransversarii muscles produce lateral flexion of the lumbar spine during unilateral activation and Nordin and Frankel (2012) stated that they perform extension of the lumbar spine during bilateral activation. However, their main functions may be proprioception (Levangie & Norkin, 2005) and spinal stability (Moore et al., 2010) due to a small cross-sectional area and short moment arm.

TYPES OF LOADS AT THE LUMBOSACRAL JOINT

There are various forces that affect the lumbosacral region, including the weight of the body itself, tension in local ligaments and muscles, intra-abdominal pressure, and any external loads such as the weight of a backpack or passenger (Hall, 2007). Each force may act along the longitudinal axis of the lumbosacral disc to create an axial or compressive load, or the force may act perpendicular to the longitudinal axis to create a shear load. Any one force may have both a compressive and a shear component, as seen in Figure 2.16, and would then be divided into these components for the purposes of force analysis. If the force acts at some distance greater than zero from the lumbosacral axis of rotation, it produces a rotation about that axis which is known as a torque or moment. As seen in Equation 1, torque (T) is the product of a force (F) and its moment arm (d⊥). The components of torque are also illustrated in Figure 2.17. A moment arm is defined as the perpendicular distance from the line of action of the force to the axis of rotation.

Equation 1: $T = F\ d_{\perp}$

where F is a force, d⊥ is its moment arm, and T is the torque it produces.

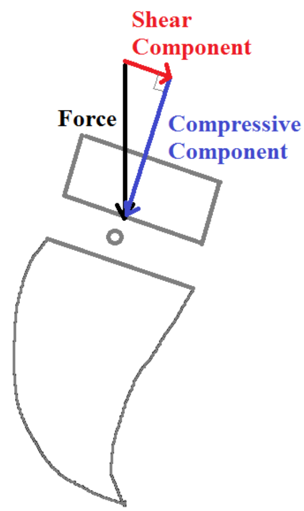


Figure 2.16: The resultant force is the vector sum of its compressive and shear components.

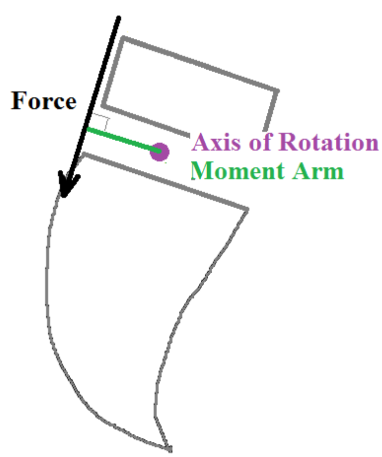


Figure 2.17: The three aspects of torque: a force, an axis of rotation, and the moment arm perpendicularly between the two.

There are various contributing factors to the compressive load on the lumbosacral region. The weight of the upper body, sometimes referred to as HAT as an abbreviation for the head, arms and trunk, is responsible for much of the axial load when upright. The portion of HAT that is supported by each vertebra increases from the superior to inferior spine. Therefore, the lumbar vertebrae experience greater compressive forces than do the cervical and thoracic vertebrae, and the fifth lumbar vertebra experiences greater compressive forces than the first lumbar vertebra. Because the centre of gravity of the upper body itself is anterior to the axis of rotation within the spinal column, the weight of HAT also creates a constant forward flexing torque that necessitates a constant backward torque produced by the posterior back muscles. The extensor muscle force used to create this torque further increases the compression at the lumbosacral joint. The ligamentum flavum and any other taut ligaments add to the axial load, as well as any stabilizing tension in the abdominal muscles. All of these factors contribute to the compressive load on the spine (Hall, 2007).

The spine is also loaded with shear forces due to body weight, load weight, muscle forces, and angle of the lumbosacral joint during standing (Cholewicki et al., 1991; Levangie & Norkin, 2005). The weight of HAT and any load carried both have a shear component which, if not resisted, would create a tendency for the upper body to slip anteriorly off the sacrum. However, shear is only a dominant force on the spine when the trunk is in a position of significant flexion (Hall, 2007). In the lumbar region, the structural elements that resist these anterior shear forces are: the inferior facet surface of one vertebra impacting on the superior facet surface of the bone below it, the orientation of the facet joints, the capsular ligaments, the annulus fibrosus of the intervertebral disc, and the iliolumbar ligaments (Levangie & Norkin, 2005). The closer the angle of the facet surfaces to the frontal plane, the more successful the resistance to the anterior shear forces; if the facet surfaces are not oriented purely in the frontal plane, the nagging shear forces may cause significant stress on the joint capsules (Levangie &

Norkin, 2005). Excessive compression on the facet surfaces and tensile stress on the joint capsules are both possible results of shear forces and they test the integrity of the joint.

As muscle forces increase, both the compression and shear forces on the vertebral joints and facet joints increase. This is because, although the line of force of the lumbar erector muscles may be simplified as parallel to the compressive force, many of the fibres of the major spine extensors are not truly parallel to the spine. However, there seems to be some debate about the resultant direction of the shear component of the muscle forces in the literature. Some sources suggest that the sum of the spinal extensor muscle forces pulls in a direction as to produce a posterior shear load in the lumbar region, which can support some of the anterior shear caused by the weight of HAT (Hall, 2007). However, many sources suggest that although this is the case at every other level of the lumbar spine, the direction of the line of pull becomes anterior at the L5-S1 level (Bogduk et al., 1992; Kingma et al., 2007; Levangie & Norkin, 2005). The magnitude and direction of application of the resultant joint shear force depends on the level of the spine, the direction of the muscle forces and force of gravity with respect to the horizontal axis of the vertebra (Hall, 2007).

The position of the trunk significantly affects the forces on the spine. When the trunk flexes, the centre of gravity of each body segment moves anteriorly which increases the length of each respective moment arm. Lengthening these moment arms increases the flexor torque caused by each weight segment; the total of this additional flexor torque necessitates an equal increase in the extensor muscle torque (Hall, 2007). Spinal extensor muscles have very short moment arms with respect to the lumbosacral flexion axis of rotation. Therefore, to resist the forward flexing torque produced by body weight and any weight carried anterior to spine, the muscles need to produce very large forces (Hall, 2007). Hall claims that these muscle forces generally make up the majority of the force acting on the spine (2007). Their tension will increase with flexion until approximately 57% hip flexion and 84% of maximum spinal flexion, when the tension suddenly ceases and the posterior spinal ligaments assume extensor responsibility (Hall, 2007). A substantial amount of shear force is exerted by the interspinous ligament when the spine is in this fully flexed position; the interspinous ligament also contributes greatly to the facet joint loading. Spinal flexion causes more compression on the lumbar spine than standing, but less than sitting (Hall, 2007). Compared to just standing, the spine is taken out of a neutral position and is compressed anteriorly and tensed posteriorly. However, the issue becomes exaggerated when progressing to sitting because the pelvis rotates backwards and the natural lordosis often disappears, allowing greater compression on the anterior disc (Hall, 2007).

Forces on the spine are also increased by acceleration. Rapid movements dramatically increase both the shear and compressive forces produced by the erector spinae. In order to minimize the maximum compression force peaks experienced at the lumbosacral joint, slow, smooth and controlled movements can be utilized.

Lumbosacral Force and Moment Values in the Literature

Over the last couple of decades, researchers have estimated the joint compressive forces, joint shear forces, muscle forces, magnitudes of forces that induce tissue failure, and different torques experienced by the lumbosacral region. Due to variations in the methods and specimens or participants used, the estimations have covered a large range of values and some have exceeded magnitudes that others had not thought possible without failure. Specifically, some of this variance may result from the use of different erector spinae moment arm lengths or differences in tissue strengths between elderly cadavers and thirty year-old elite weight lifters. Also, individual differences in lifting strategies can result in slightly different positions and therefore slightly different proportions of load at the knee, hip, and lumbar spine (Cholewicki et al., 1991). Force estimations within the same specimen may even vary when using different methods, such as electromyography, inverse dynamics, static equilibrium analysis, or in vivo disc pressure measurements. Because many models that discuss forces in the lumbar region have focussed on the L4-L5 joint, data has been included here for both the L4-L5 and L5-S1 joints. Because the slope of the L4-L5 joint is less than that of the L5-S1 joint, its joint compression values may be higher and its shear joint force values may be lower than what would be typically experienced at the L5-S1 joint. Acknowledging these inconsistencies, example values are provided in the following paragraphs for several types of forces and torques.

Lumbosacral Compressive Joint Reaction Forces

Some researchers have estimated the compressive forces experienced at the lumbar spine while standing. In anatomical position, the lumbosacral joint experiences peak compressive loads of 0.82 to 1.18 times the weight of the total body (Levangie & Norkin, 2005). During walking, these loads increase to 1.41 to 2.07 (Levangie & Norkin, 2005) or 1.2 – 2.5 times body weight (Cappozzo et al., 1984; Khoo et al. (1995); Goh et al, 1998). Khoo et al. (1995) found the mean of lumbosacral joint compression while walking to be at 1.71±0.27 times body weight, ranging from 1.44 to 2.07 times body weight. The results presented by Khoo et al. likely under-represent the actual loads, as they were under the impression that the abdominal muscles off-set the load on the spine (1995), while current research suggests otherwise. It will be interesting to determine how these values increase further when backpacking and piggybacking with a moderate load.

The forces and moments at the lumbosacral joint increase as the spine is asked to support more load (Goh et al., 1998). The average lumbosacral force was 1.5±0.24, 1.90±0.27, and

2.46±0.29 times body weight when walking unloaded, with a load of 15% body weight and with a load of 30% body weight, respectively (Goh et al., 1998). Adding these 15% and 30% body weight backpacks increased the lumbosacral force by 27% and 64% (Goh, Thambyah, & Bose, 1998). While walking with 5 kg or 30 kg in each hand, the L4-L5 compression forces were 1225N and 2339N respectively; when the 30 kg load was carried in only one hand the compression forces increased to 2874 N (McGill et al., 2013). When standing upright or walking, the abdominal and back muscles tend to co-contract to give the trunk stability. However, the total compressive load on the L4-L5 joint can be further increased by up to 1000 N, usually 300 N to 700 N, when the trunk muscles are purposefully co-contracted to stabilize the trunk during an activity (Kavcic et al., 2004). Therefore, contracting the musculature to create an anatomical abdominal belt not only increases the stability, but also the total compressive load on the spine (Kavcic et al., 2004).

In comparison to unloaded standing and walking, much greater forces are experienced in the lumbar spine while performing other activities; some examples are listed in Table 2.1. Researchers have found normal compressive forces at the lumbosacral joint to range up to around 7000N. During a single-leg extension exercise with a relatively neutral spine in a hands and knees position, expected compression forces are approximately 2018N (Kavcic, 2004) to 2500 N (Levangie & Norkin, 2005). While in this four point position, contralateral arm and leg extension has been seen to increase the joint compression to approximately 2740N (Kavcic, 2004). These values were considered to be relatively low and can be compared to 3300N compression forces during sit-ups or curl-ups, or 6000N compression forces during simultaneous arm and leg raises from a prone position (Levangie & Norkin, 2005). McGill (1995) estimated the L4-L5 joint compression load at 3234N during the isometric straight leg sit-up, and also over

3000 N for bent leg sit-ups. Sitting on a stability ball, on a chair, performing a bridge position, performing a bridge with a single leg-extension, and performing a side-bridge exhibited L4-L5 compression forces of 2097N, 2128N, 2387N, 2707N, and 2726 N of joint compressive forces, respectively (Kavcic, 2004). Compressive forces ranged from 490 to 2660 N throughout a variety of activities tested by Schultz and colleagues (Schultz, Andersson, Hadespeck, Ortengren, Nordin, & Bjork, 1982; Schultz, Andersson, Ortengren, Bjork, & Nordin, 1982; Schultz, Andersson, Ortengren, Hadespeck, & Nachemson, 1982); these values were less than the 2811 N extensor maximum compressive force predicted by Bogduk et al. (1992). While lifting 18kg, compressive forces were estimated at an average maximum of 5218N to 6391 N at the L4-L5 disc, which has a lesser angle of inclination than L5-S1 by approximately ten degrees (McGill & Norman, 1985). Because this was an average maximal value, there were inherently values both higher and lower than these listed. Compression at the L4-L5 joint was 5647 N with a 32 kg load during squats (Potvin et al., 1991).

Table 2.1: Compressive and Shear Joint Reaction Forces during Various Strenuous Activities

	Load (kg)	Activity (T)	L4-L5 Fc (N)	L4-L5 Fs (N)
McGill & Norman 1986	27.3	449	7296	612
(Ab Balloon Theory)	54.5	331	5571	304
	90.9	431	8921	843
McGill & Norman, 1985	18		5218-6391	
Potvin et al., 1991	32	Squat Lifts	5647	
Cholewicki et a.l, 1991	145.8	Female Power Deadlift	5844-7077	1107
	256.7	Male Power Deadlift	10290-17192	1739
McGill et al., 2009	38-180	Strongman Events	6890-12043	913-2409
Lander et al., 1990		Parallel Squat		3843

To understand the relative risk of certain magnitudes of compression forces, it is important to address the magnitudes of both maximal forces experienced by elite athletes and forces that produced tissue failure. The athletes that experience the greatest loads on the spine are likely weightlifters and powerlifters, some experienced compressive joint reaction forces up to over 17 192N (Cholewicki, McGill, & Norman, 1991). While performing deadlifts averaging 145.8±18.4 kg, female elite power-lifters experienced 5844 N to 7077 N of compressive force at the L4-L5 disc (Cholewicki et al., 1991). Their male counterparts lifted an average of 256.7±29.9 kg with averages ranging from 10290 N to 17192 N of compressive forces at the L4-L5 disc (Cholewicki et al., 1991). The authors suggest that this is beyond any documented tolerance limits, although they do not state the exact limits to which they are referring (Cholewicki et al., 1991). Maximum lumbosacral forces have been estimated at 10473 N for the parallel squat (Lander et al., 1990). During walking Strongman events with average loads from 38 kg to 180 kg, such as the farmer's walk or keg walk, the average L4-L5 joint compressive force ranged from 6890 N to 12043 N (McGill et al., 2009). Compressive forces have also been seen to surpass 12043 N in competitive Strongmen (McGill et al.b, 2009). Because of the frequency of high magnitude loads on the spines of these athletes, it is expected that their tissues would have responded and gained in strength. Therefore, it is not surprising to see that they may regularly experience greater forces than forces that rupture cadaveric lumbar spine specimens (Cholewicki et al., 1991).

There does not seem to be an accepted value for lumbar tissue failure, although the weakest tissue in compression is known to be the vertebral endplate (Cholewicki, McGill, & Norman, 1991). Without the muscles producing tension to stiffen the spine, the lumbar spine has been seen to buckle under as little as four Newtons of force (Hall, 2007). Also without the help of muscles, cadaverous thoracolumbar spines have been seen to buckle at only 20 to 40 N (Nordin & Frankel, 2012). This critical load is much greater in vivo, and will depend on each unique body. The lumbar tissues of elite weight lifters would be expected to be stronger than the tissues of healthy young participants and both would be expected to be stronger than the tissues of elderly cadaveric specimens that may have likely experienced atrophy. Cholewicki et al. (1996) found that isolated motion segments of the lumbar spine failed at approximately 90 N squared. Jager (1987) found that lumbar cadaver motion segments failed between around 2000N and 8000N, depending on the study, averaging around 5000 N across twenty five studies (Chaffin et al., 1999). Benedek and Villars (1974) suggested a much greater value; the wet lumbar intervertebral disc was elastic until approximately 9810N, but ruptured at approximately 14715N or when compressed to 35% of its height (Benedek & Villars, 1974). This failure load increased from 3139N in the cervical vertebrae to 44145N in the upper thoracic, to 11282N in the lower thoracic and 14715N in the lumbar spine, according to the amount of surface area supporting the load (Benedek & Villars, 1974). These higher values are more logical based on the common and seemingly harmless loads experienced by powerlifters, which can be nearing 18000 N (Cholewicki et al., 1991). During near-maximal load deadlifts into full trunk flexion, Cholewicki et al. (1991) were not able to identify any quantifiable shearing or compression of the disc by means of videofluroscopy (Cholewicki et al., 1991).

There have been a few suggestions for maximal lift limits. McGill (1995) stated that the lumbar spines of some individuals would even experience substantial risk during compression forces over 3000 N. The National Institute for Occupational Safety and Health (NIOSH) recommended limits for compressive joint reaction forces in the workplace. In 1981, they recommended that the average worker should avoid joint compressive forces greater than 3433 N, and that the maximal permissible limit was 6377 N (NIOSH, 1981).

Lumbosacral Shear Joint Reaction Forces

Researchers have also attempted to quantify lumbar shear forces, which have been seen to surpass 2409 N (McGill et al.b, 2009), and have been documented up to 3843 N. Lumbosacral joint shear was 0.22±0.24 times body weight, ranging from 0.06 to 0.63 times body weight while walking unloaded (Khoo et al., 1995). Joint shear forces at L4-L5 were 167±113.2 N when carrying 5 kg in each hand, 294±102.1 N when carrying 30 kg in each hand, and 519±224.3 N when carrying 30 kg in one hand (McGill et al., 2013). The L4-L5 joint anterior shear force was 257 N during the straight leg sit-up (McGill, 1995). The joint shear force was 1080 N at the L3-L4 intervertebral joint while maintaining an isometric moment of 50 Nm (Kingma et al., 2007). The L4-L5 shear force reached 760 N during a study on the overhead cable press (McGill et al.a, 2009). Joint shear reached 483.2 N at the L4-L5 joint with a 22 kg load during stoop lifts (Potvin et al., 1991). Previous studies using various loads and lifts showed shear forces under 300 N at the L4-L5 joint (Kingma, Bosch, Bruins, & van Dieen, 2004; Potvin, Norman, & McGill, 1991) but they ranged from around 600 N to 1500 N at the L5-S1 joint during lifting (Kingma & van Dieen, 2004; Kingma et al., 2007; 18,Kingma et al., 2004; Mandell, Lipton, Bernstein, Kucera, & Kampner, 1989; Marras & Davis, 1998).

It is also important to look at the shear forces during maximal lifts, in order to understand the risk and tissue limits inherent in these activities. Female power-lifters exhibited a mean joint shear force of 1107±200 N and male power-lifters exhibited a mean joint shear force of 1739±286 N when performing a deadlift (Cholewicki et al., 1991). The average L4-L5 joint shear force ranged from 913 N to 2409 N during walking Strongman events for average loads ranging from 38 to 180 kg (McGill et al.b, 2009). Lander et al. (1990) estimated maximum joint shear forces to be approximately 3843 N during the parallel squat. Just like the uncertainty for compression forces, the maximum tolerances for shear joint forces in the lumbar spine are not concrete. The pars interarticularis has been seen to fracture at joint shear loads of approximately 600 to 4000 N (Cyron, Hutton, & Troup, 1976; Lamy, Bazergui, Kraus, & Farfan, 1975).

Lumbosacral Resultant Joint Reaction Forces

The compressive and shear reaction forces are the two components of the total joint reaction force. The compressive/shear force ratios were 1.48/0.28, 1.87/0.35, and 2.42/0.45 times body weight with respect to the load conditions from unloaded to15% to 30% body weight (Goh et al., 1998). When walking with loaded backpacks of 15% and 30% body weight, the shear forces were generally less than the compression forces at the lumbosacral joint by approximately 81±8% (Goh et al., 1998). While walking unloaded, shear was approximately 3% to 32% and compression was approximately 95% to 100% (Khoo et al., 1995). Total joint reaction force has been seen to be 1.73±0.28 times body weight (Khoo et al., 1995). When walking, peak resultant joint loads were greater at the final peak compared to the initial ground reaction force peak, with the peak lumbosacral load occurring between 68% and 91% of stance phase (81±8%) (Khoo et al., 1995). Khoo et al. (1995) speculated that the loads during single stance phase were likely the greatest loads on the spine because the upper body weight and the spine are supported by only one limb.

Moments at the Lumbosacral Joint

It is also useful to identify the range of estimated moments about the mediolateral axis of rotation of the lumbar spine. These moments are produced by muscle forces, the load, and the weight of the portions of body superior to the lumbosacral joint. An example of a very strong moment acting about a mediolateral axis of rotation in the lumbar spine would be 1100Nm, which would only be estimated for an elite athlete. The estimated torque sustained by the fifth lumbar vertebra during standing was approximately 230 Nm in healthy 30 year old males, 189 Nm in males over age 30, and 102 Nm in females over age 30 (McNeill et al., 1980). The L4-L5 average moment ranged from 61 to 267 Nm while walking and carrying heavy loads in various positions (McGill et al.b, 2009). The estimated maximum lumbar torques using three models ranged from 209 Nm to 539 Nm while lifting an 18kg load at an extreme reach of 0.83 meters (McGill & Norman, 1985). The average L4-L5 moment for an 18kg lift at a reach position of 0.83 m was 232 Nm calculated from the static model and 276 Nm calculated from the dynamic model (McGill & Norman, 1985). Respectively, these average moments translated to 5218 N and 6391 N of compression force on the L4-L5 disc (McGill & Norman, 1985). Maximum lumbosacral moments have been estimated at 669 Nm for the parallel squat (Lander et al., 1990). When deadlifting, elite athletes may experience torques ranging from 254 Nm to 460 Nm for women and 445 Nm to 1071 Nm for men (Cholewicki et al., 1991); the highest average moment was 988 Nm. During a 309 kg tire-flip strongman event, the average moment was 792±58 (McGill et al.b, 2009). An elastic limit of 70 Nm was suggested for the motion segments (Miller et al., 1986), and failure of the motion segment was seen at 156 Nm by Osvalder et al. (1990), although it is obvious that value can be surpassed.

Forces and Torques Produced by the Extensor Musculature

The muscle forces experienced during these types of activities have also been documented. Farfan (1973) estimated the average maximal force able to be produced by the erector spinae to be between 2200 and 5500 Newtons (Chaffin et al., 1999). The mean maximal voluntary lumbar extensor muscle force in a modified kneeling position with a neutral spine was 345.4±126.7 N for fifteen men and women with a mass of 72.2±14.6 kg (Russ et al, 2009). While the thoracic fibres of the erector spinae result in the majority of the compression load on the upper lumbar segments, the compression is increasingly caused by lumbar fibres of the erector spinae and multifidus toward L5-S1 (Bogduk et al., 1992). The thoracic and lumbar erector spinae muscles each contributed approximately 995 and 1190 N of compression force at the L5-S1 joint when in the upright position (Bogduk et al., 1992). When the multifidus was included, the total compression force at the L5-S1 joint increased to 2799 N, as the multifidus contributed 614 N of compression force at this joint (Bogduk et al., 1992). The specimens used in the study by Bogduk et al. in 1992 were elderly cadavers and therefore may have produced slightly different torques as compared to 30 year old lumbar spines in vivo, for example. The average muscle compression force reached 4185 N during cable-walkouts with 5.4±1.6kg of load (McGill et al.a, 2009). During walking Strongman events with average loads from 38 kg to 180 kg, such as the farmer's walk or keg walk, the average compressive muscular forces ranged from 5492 N to 8020 N (McGill et al., 2009).

The average L4-L5 muscular shear force ranged from 1598 N to 2519 N during Strongman events with average loads ranging from 38 to 180 kg (McGill et al.b, 2009). The thoracic fascicles of the erector spine are essentially parallel to the spine and therefore exert minimal shear forces on the lumbar spine, some in the posterior and some in the anterior direction. However, when the effects of the many fascicles of this muscle are summed, the thoracic fascicles of the erector spinae produce a major proportion of the shear forces exerted on the lumbar vertebrae (Bogduk et al., 1992). At the L5-S1 level, the total net shear produced by all of the erector spinae and multifidus is an anterior shear of -7452 N (Bogduk et al., 1992).

Lumbosacral extensor muscle moments ranged from 144 Nm to 450 Nm. Bogduk et al. (1992) was one research team that calculated these values as exerted by each individual fascicle at each individual lumbar segment while standing. This was done in much greater detail than the scope of the present study; however, the values provide an estimate as to the magnitude of forces occurring in each aspect of the lumbar spine while a person is standing with no load. At different levels of the lumbar spine, the muscles provide different proportions of the extensor moment. At both the L4-L5 and L5-S1 joints, the extensor torque provided by the lumbar fascicles of multifidus and lumbar fascicles of the erector spinae together equal approximately the magnitude of extensor torque provided by the thoracic fascicles of the lumbar erector spinae (Bogduk et al., 1992). The lumbar erector spinae and the thoracic erector spinae bilaterally generate approximately 44 Nm and 69 Nm respectively at the L5-S1 joint in the upright position (Bogduk et al., 1992). The total extensor moment produced by the multifidus muscle was approximately 31 Nm at L5-S1 (Bogduk et al., 1992). Including extensor moments produced by the multifidus, the total bilateral moment produced was 144 Nm (Bogduk et al., 1992). These torque calculations are based on the size of the muscle and the amount of force that is able to be produced by a certain cross sectional area of muscle in that region. Bogduk et al. (1992) assumed that the muscle in the lumbar region can produce $46Ncm^2$. McGill and Norman estimated the maximum extensor torque to be 450 Nm at the L4-L5 segment (1986); Bogduk et al. (1992) later suggested that they overestimated the size of the musculature in their calculations and therefore overestimated the total torque.

Total Extensor Torque- Lumbar Structures Potentially Capable of Trunk Extension

Typically, individuals tend to lift loads in a position anterior to the trunk, with the centre of gravity of the load anterior to the mediolateral axis of rotation of the lumbosacral joint. With a heavy load, the trunk may experience such great flexion due to such a great flexor torque that all of the structures that could possibly exert a trunk extension moment are summoned. These structures are not expected to experience such extreme stresses in the less extreme positions that are expected during the posterior load carriage in this study. However, it is important to understand which tissues are capable of assisting with lumbar extension. In addition to any muscles, the portion of each ligamentous, fibrous, and fascial structure that is posterior to the mediolateral axis of rotation of the lumbosacral joint may contribute to the total extensor torque at that joint. The collagen fibres in each tissue help maintain their structure when stressed, and will therefore produce an extensor torque when subjected to the posterior tension caused by substantial spinal flexion.

Besides the trunk extensor muscles, the other structures in a position to contribute to an extensor torque are: 1) the posterior layers of the annulus fibrosis, 2) the ligamentum flavum, 3) the posterior facet joint capsule, 4) the posterior longitudinal ligament, 5) the interspinous ligaments, 6) the supraspinous ligaments, and 7) the thoracolumbar fascia. The length of the moment arm of the extensor force increases from the posterior longitudinal ligament to the ligamentum flavum to the supraspinous ligament. Bogduk (1989) suggested that the possible forces and moment arms of the torques exerted by each structure respectively are as follows: 1) 500 N x 2 cm = 10 Nm; 2) 1000 N x 3 cm = 30 Nm; 3) 900 N x 8 cm = 72 Nm; 4) 300 N x 2.5 cm 7.5 Nm; 5) 200 N x 5 cm = 10 Nm; 6) 200 N x 6 cm = 12 Nm; 7) 335 N x 6.5 cm = 20 Nm (Bogduk 1989). These values are seen in Table 2.2 below, along with some other values for comparison. When Bogduk (1989) included the erector spinae muscle force of 2500 N at 10 cm from the axis, this extra 250 Nm of torque made the estimated total possible extensor torque increase to slightly greater than 400 Nm for this 50 kg example load. This contradicts predictions made by McGill and Norman (1986), which suggested that the muscles alone could contribute 450 Nm. If any of these structures were injured, the maximum load would decrease.

Table 2.2: Forces, Moment Arms, and Torque for Potential Lumbar Extensor Tissues

Posterior Structure	Force (N)	Moment Arm (cm)	Torque (Nm)
[1]Layers of annulus fibrosis	500	2	10
[1]Ligamentum flavum	1000	3	30
[4]Ligamentum Flavum	3 (only Fc)		0
[1]Facet joint capsule	900	8	72
[4]Capsular Ligaments	0		0
[1]Posterior longitudinal ligament	300	2.5	7.5
[1]Interspinous ligaments	200	5	10
[4]Interspinous	29	3.5	1
[1]Supraspinous ligaments	200	6	12
[4]Supraspinous	387	7	27
[1]Thoracolumbar fascia	335	6.5	20
[3]Lumbodorsal Fascia	61-238	~7.5	4.6-17.6
[4]Lumbodorsal Fascia	101.5	6.8	7
[1]Erector Spinae Musculature	2500	10	250
[2]L4-L5 Erector Spinae	3358	11.8	284
[4]Erector Spinae	4497	6.8	307

[1] Bogduk, 1989
[2]McGill & Norman, 1986
[3]McGill & Norman, 1988
[4] Potvin et al., 1991

Several other researchers have tried to define the role of these passive structures in trunk extension and have quoted lower values. Cholewicki and McGill estimated that the posterior ligaments could contribute approximately 30 to 40 Nm of extensor moment during near-maximal deadlifts (Cholewicki & McGill, 1991). Estimations by Potvin et al. (1991) agreed that this value was typically no more than 60 Nm for a lifting activity, demonstrating a range from 3.4 to 52.4 Nm during stoop and squat lifts with loads of 6 to 32 kg (Potvin et al., 1991). However, in a fully flexed position with no muscular support, they may reach about 80 Nm (Potvin et al., 1991). They provide a greater proportion of the extensor moment with increased trunk flexion, possibly not contributing until approximately six degrees less than full trunk flexion (Potvin et al., 1991), and then still only at negligible amounts (McGill & Norman, 1986). With five kilogram loads, the posterior ligaments provided less than 16 % of the total moment, compared to peak muscle moments of 170 to 331 Nm in the same study (Potvin et al., 1991). The role of the lumbodorsal fascia is likely even less; Potvin et al. (1991) state that its contribution to the total moment is negligible, and McGill and Norman (1988) note a range from 1 to 4 % of the total extensor moment during squat lifts. Overall, it seems that the posterior ligaments do not provide substantial extensor torques, but they ultimately limit the end range of motion of trunk flexion (Cholewicki & McGill, 1991; McGill & Norman, 1986).

TYPICAL LUMBAR INJURIES

As it may be evident, the lumbar spine experiences many high muscle forces over the course of even one day. It is also subjected to the weight of gravity of every more superior portion of the body, the weight of any loads being carried, and any other external forces. The structures of the lumbar spine are designed to withstand these forces and even often adapt to become stronger over time, but may succumb to injury if the volume or magnitude of stress becomes too great. As previously noted, pain or injury of the back is one of the most common chronic conditions in Canada (Schultz & Kopec, 2003).

There are three common categories of activities that have a high incidence of back injury: activities that demand either frequent weight-loading, rotation, or hyper-extension (Alexander, 1985). Weight loading activities subject the spine to high volumes of compression forces, rotational activities subject the spine to a forceful twist, and hyper-extension activities often repeatedly and quickly take the spine into and out of an over-arched and compromised position. The piggyback involves not only weight-loading but also possible rotation during loading, off-loading, and when controlling the movements of the load. Manual materials handling, which is well-documented and involves lifts and carries like piggybacking, is considered a high risk activity in the workforce (Health and Safety Executive, 2007). Because any weight anterior to the lumbosacral axis of the spine increases the flexion moment, the erector spinae muscles resisting trunk flexion often ache if the load is carried for an extended period of time (Moore et al., 2010).

Most often, cases of lumbar back pain will involve soft tissue injuries such as contusions, muscle strains, or ligament sprains resulting from an impact of overloading of the muscle (Alexander, 1985; Hall, 2007). Lumbar injuries may also involve different types of bone fractures. Injury to the bone will only briefly be discussed in the next section because piggybacking has not been suggested as a common mechanism of injury for fracture. Lumbar issues due to stress or carrying loads often involves injury to the intervertebral disc, which can also be injured via acute trauma. The mechanism of injury to the disc usually involves the high magnitudes of anterior shear force while in trunk flexion (Alexander, 1985). The vertebral endplate may separate from the intervertebral disc, creating fissures in the annulus and allowing some of the nucleus pulposus to leave its normal position within the annulus. Because trunk flexion produces compressive forces on the anterior disc and tension on the posterior disc, the nucleus pulposus often breaks through the annulus fibrosus and exits the posterior or lateral posterior corner of the disc (Alexander, 1985; Hall, 2007). This is known as a disc herniation.

As seen in Figure 2.18, this displaced material presses on the adjacent spinal cord or nerves to create pain or numbness. Sensation may also be detected by sensory nerves in the posterior longitudinal ligament, the vertebral body and the articular cartilage of the facet joints if they are affected. However, the disc itself is not innervated. The L4-L5 disc and the L5-S1 disc, as well as cervical discs, are most frequently herniated (Hall, 2007). Most cases of lumbar herniation are not operated on, but may receive treatments such as physiotherapy, medication, lumbar injections, or lumbar discectomy (Hall, 2007). Although disc herniations are produced and/or aggravated by loaded trunk flexion, the trunk flexion angle is not expected to be excessive while piggybacking a much lighter participant. The greater the weight of the person being carried, the greater the expected trunk flexion angle and the greater the risk of disc herniation.

Figure 2.18: Herniated lumbar disc. A portion of the nucleus pulposus has exited through a fissure in the annulus fibrosus and is pressing on the nerve tissue.

Specific Vulnerability of the Lumbosacral Joint

The lumbar region is the most injury-susceptible portion of the spine because of its location, structure, and the quick movements and high forces to which is it often subjected (Alexander, 1985). The lumbar spine is the only rigid structure connecting the upper body to the lower body, and each vertebra carries the weight from the portion of the body above it (Alexander, 1985). For this reason, the most inferior motion segment, the L5-S1 segment, is exposed to the greatest loads. The lumbar and cervical regions are most mobile and are therefore most vulnerable to injury (Moore et al., 2010); the lumbar region pairs the greatest loads with the most mobility, and a structure designed for increased mobility usually has decreased stability. The lumbosacral joint also experiences the greatest moment arm when a load is being lifted in front of the body (Chaffin et al., 1999).

The L5-S1 disc is injured the most often, whether the injury is acute, degenerative, or if the disc has prolapsed (Hall, 2007; Khoo et al., 1995). The superior sacrum, the resting surface for this fibrocartilage disc, is at a much greater angle than is the disc at any other level of the spine. As discussed earlier, this angle causes a high proportion of shear forces when loaded. These shear forces place stress on fibres of the annulus fibrosus via tensile forces and on the neural arch via compression forces at the facet joints. Carrying a load further increases both the shear and compression forces.

The lumbosacral joint is also under constant stress caused by the erector spinae muscle group as they maintain an upright posture. The centre of mass of the upright human body is anterior to the spine and therefore anterior to the mediolateral lumbosacral axis of rotation (Alexander, 1985). The force of gravity acting on the centre of mass therefore results in a constant forward moment, which is resisted by continuous trunk extensor contractions of relatively low magnitude. However, the moment arm of the weight is longer than the moment arm of the muscle force so the muscle must exert a force greater than that of the gravity and the total force required over time is substantial (Alexander, 1985). The erector spinae group attaches to both the transverse and spinous processes of the lumbar vertebrae, and therefore exerts substantial tensile forces on the vertebral arch. Over time, this constant stress on the vertebra can cause it to fracture at its weakest portion, the vertebral arch, leading to either a unilateral or bilateral fracture of the pars interarticularis. The pars interarticularis is the portion of bone between the superior and inferior articular processes; a unilateral fracture is known as spondylolysis and a bilateral fracture is known as spondylolisthesis (Bogduk, 1987).

The facet joints of the vertebral arch resist anterior sliding of each vertebra on the vertebra below it. Both the comparatively high shear forces and the tensile muscle forces at the vertebral arch increase the risk of a neural arch fracture at the lumbosacral level specifically. The extreme angle of the base of the sacrum increases the risk of neural arch fracture, and when the neural arch or facet joint structures are compromised, this angle increases the likelihood that the vertebral body will slip anteriorly (Nordin & Frankel, 2012). Trunk flexion, which is induced by load carriage, theoretically increases the length of the moment arm of the force of gravity, which would increase the magnitude of the shear force and the erector torque necessary to maintain an upright posture. The shear force itself would be increased by an increase in the sacral angle, anterior pelvic tilt, load, and or the increased erector spinae torque caused by a load. Therefore, carrying load is likely to increase the risk of spondylolysis and/or spondylolisthesis at the lumbosacral joint. The anterior and posterior longitudinal ligaments, the ligamentum flavum, and neural arch also experience a greater shear stress when the lumbosacral angle is increased (Alexander, Adv Bio, 2010).

For the reasons mentioned above, the lumbosacral joint is the articulation that will be most important in evaluating the stress to the lower back when carrying a load in this trunk-flexed load-carrying position.

LITERATURE ON LOAD CARRIAGE

Scientific research on carrying another person is lacking, and no studies on the piggyback have been found. Only a small variety of exercise or life articles in newspapers and magazines mentioned piggybacking. However, the scientific literature on carrying a loaded backpack is extensive. It has been investigated with respect to school children, recreational hikers, and military personnel, questioning the effect of the position of the pack, the weight of the pack, safety, et cetera. Many load carriage studies addressed physiological factors, but the outcomes of biomechanical studies will be the focus of this literature review. Of these, some addressed the effect of loading on gait parameters and a much smaller portion addressed the forces affecting the lumbar spine. An article published in March, 2013 claimed that more research should investigate kinematic changes at the lumbar spine due to load in order for appropriate military loading guidelines to be set (Rodriguez-Soto, 2013).

PIGGYBACKING

Piggyback rides can be seen anywhere and have been documented by countless non-scientific sources. One boot-camp noted a two hour partner carry “up and down the quay” on their facebook page (Piggyback New West, March 13, 2013). Photos on the world wide web prove that this activity has been used in sport team training for at least several professional teams, including the Wales’ National Soccer Team (Associated Newspapers, 2011), the Nottingham Rugby Team (Nottingham Rugby, 2011), and the American Longhorn Football Team (Payson Roundup, 2013). Included as part of a football training program, Palmieri claimed that running with someone on your back improved trunk lean when posture was too upright (1993). It was included in athletic races such as the Lil Mudder (Madliger, 2012) and Tough Mudder (Muller, 2013). In the Edmonton Journal, a mother wrote in to Dr. Donohue asking if it was safe for her daughter’s high school soccer team to train by carrying each other on the back (Donohue, 1996). The doctor wrote that this exercise was not advised as it involved a lot of shifting dead weight carried on the back, and that it would possibly disrupt natural stride and balance to increase the risk of injury to the back and legs. A fitness article in the Washington Post suggested performing squats while piggybacking as part of a romantic partner exercise routine (Schneider, 2009). Between squats, the carrier was instructed to walk ten to twenty feet. When participating in this exercise, one partner would likely be significantly heavier than the other. Without previous training, this may be a risky activity for the person that weighed less, but the author does state any possibility of risk (Schneider, 2009).

Not only is the piggyback carry used specifically for strength training, but it is also used to promote bonding while being physically active. The coach of a professional soccer team in the United Kingdom used the partner carry to distract his players from the ensuing playoffs and promote “togetherness” (Nottingham Rugby, 2011). A few sources noted that enriched parenting and child development also result from this activity. An article in the Ottawa Citizen encouraged piggybacking as an important aspect of natural rough-and-tumble play, which increased physical literacy and educated children on leadership, dominance, co-operation and group social skills (Bennett, 1999). Bennett also stated that piggybacking was very important between parents and children (1999). In a 2007 newspaper article, a father noted the physical difficulties of carrying a child on the back as it became painful as an older parent (Lileks, 2007). Because this activity was recognized as a normal part of parenting and recreational play, it was still performed by the parent and the discomfort was ignored (Lileks, 2007). The YMCA had a father-daughter day that began with a piggyback race (YMCA, 2012), which would also promote a healthy family relationship. Together, these articles illustrate the interest in and prevalence of piggybacking as a recreational activity. This research will not only provide knowledge about the biomechanics or potential risk during competition, it will also provide insight applicable to piggybacking in life.

Piggybacking in CrossFit

Although individuals have been carried on the backs of parents, siblings, teammates, boot-camp cohorts, and friends for decades, this activity has more recently been introduced into competition through the relatively new sport of CrossFit. CrossFit has been quickly gaining popularity in the fitness community. There are currently six facilities that practice CrossFit in Winnipeg; at least three that occasionally use the piggyback as a training tool. An example of such training is seen in Figure 2.19. The Winnipeg facilities were contacted via email. One facility replied that they had only been open for four months and had not expanded programming to the piggyback yet (CrossFit Sublime, 2013), and two facilities did not reply. The Winnipeg CrossFit facilities that have trained with the piggyback carry use it occasionally, usually when hosting a team or partner session, and the frequency of this drill increases before competitions (CrossFit Winnipeg, 2013; CrossFit 204, 2013; Prairie CrossFit, 2013). They do not have guidelines for the recommended relative weight of each partner and participants will often carry someone heavier than themselves, but the coaches seem to be mindful of the notion that they should evaluate strength and weight when pairing partners (CrossFit Winnipeg, 2013; CrossFit 204, 2013; Prairie CrossFit, 2013). In competition, the teams tend to be instructed to have each participant carried for a certain distance. This means that they are often able to arrange the most suitable partner pairs, although they may not always be able to avoid an intensely challenging carry due to relative partner weights (CrossFit Winnipeg, 2013; CrossFit 204, 2013). When investigating photos of the piggyback carry in CrossFit on the world wide web, it seems that this activity has been used extensively throughout the CrossFit community. It is evident that not only do athletes need to carry another person on the back while walking, they also piggyback while

Figure 2.19: FrostFit 2014. An example of piggyback training at a Winnipeg CrossFit competition (Skrypnyk, 2014).

running, racing, squatting, and combinations of these exercises. In order to better understand the biomechanical demands of such a task, this study will directly investigate the kinetics and kinematics of piggybacking, while considering the results of previous studies on backpack carriage.

STUDIES ON BACKPACK CARRIAGE

Several previous studies that involved backpack carriage will be outlined here, to provide an idea of the type of research that has been performed. Biomechanical backpack studies addressed load carriage and one or more of several factors: trunk position, other kinematic parameters, temporal parameters, ground reaction force values, and/or forces at the L4-L5 or L5-S1 joint.

A 1999 study by Vacheron, Poumarat, Chandezon, and Vanneuville investigated trunk positions when unloaded and loaded with a 22.5 kg backpack. They had participants of three levels of mountain-hiking experience carry a load within a backpack at the level of the ninth thoracic vertebra, walking with the empty backpack and then with the loaded backpack (Vacheron et al., 1999). They used a Vicon system with various markers on the spine to compare the positions of different segments of the spine, and found that trunk flexion occurred while carrying the backpack. Vacheron et al. (1999) also found that the S1-L3-T12 segment decreased range of movement during gait when loaded.

Al-Khabbaz, Shimada, & Hasegawa (2008) performed another related backpack study, although their methods only involved standing and the results varied from the results of studies with locomotion. They compared changes in trunk flexion angle while standing with no load, a load that weighed 10% body weight (BW), a load that weighed 15% BW, and a load that weighed 20% BW. These researchers used a Vicon system to measure trunk flexion, side flexion, and rotation, and surface electromyography (EMG) to measure the relative strength of trunk muscle contractions. Nineteen participants performed two trials per condition (Al-Khabbaz et al., 2008). They used unloaded standing trunk posture as a 0 degree reference. Each additional 5% BW load further extended the standing trunk posture by around three degrees; with backpack loads of 10%, 15%, and 20% BW, the trunk extended a further 3.37, 3.02, and 3.9 degrees (Al-Khabbaz et al., 2008). It is interesting that they found that the trunk extended when most research has found that the trunk flexed with the addition of a posterior load. It is also interesting that they found the abdominal muscles to increase in strength of muscle contraction with added load while the erector spinae did not.

However, the fact that they found the trunk to extend explains this EMG result; with no compensatory trunk flexion, the load was creating a larger extensor torque, which needed to be balanced by an abdominal flexor torque. They suggested that the strange EMG result may have been because the erector spinae were already activated to hold the body in an upright posture, and then the abdominals also started to contribute when the backpack load was added (Al-Khabbaz et al., 2008). Because the centre of gravity of the upper body is known to be anterior to the lumbosacral joint, the erector spinae are already known to be active during upright posture (Al-Khabbaz et al., 2008; Alexander, 1985). It seems that this study explains the initial torque changes with load, but the body may instinctively begin to flex the trunk forward when needing to walk. Because this study was only performed in a standing position, it can provide some insight but is not an accurate representation of walking with a loaded backpack.

A 2007 military backpack study by Birrell, Hooper and Haslam addressed the effect of load carriage on ground reaction force patterns. Military backpacks may be worn alone, or they may be worn in combination with webbing, which distributes load into various pouches surrounding the trunk. The webbing is formally called personal load carrying equipment and is typically composed of a belt, shoulder harness, and several pouches. Birrell et al. (2007) compared eight conditions of load carriage: 1) unloaded, 2) rifle, 3) rifle and 8kg load in webbing, 4) rifle, 16 kg in webbing, 5) rifle and 16 kg in backpack 6) rifle, 8kg in webbing and 16 kg in backpack, 7) rifle, 16 kg in webbing and 16 backpack, 8) rifle, 16 kg in webbing and 24 in backpack. A force plate was used to measure the ground reaction forces and the article does not state how the kinematics were measured. They found that each eight kilogram load increase resulted in a proportional increase in the vertical ground reaction force, which is logical (Birrell et al., 2007). When 16 kg of load was carried in the more centralized webbing rather than the posterior backpack, there were greater vertical ground reaction force impact peaks and the stance duration was shorter; the values of these results were shifting towards that of the unloaded condition (Birrell et al., 2007). Much like with the webbing, the mass is more centralized during piggybacking than backpacking. Birrell et al. (2007) suggested that the greater impact peak experienced when carrying the load in the webbing was because of a greater proportion of the load was actually under the foot at impact, which is likely because of a more natural gait pattern in comparison to backpack carriage.

The work that is most related to the concepts discussed in this study was executed by Goh, Thambyah, and Bose in 1998. They compared the peak lumbosacral spine forces and a few kinematic and gait parameters between three load conditions: no load, a backpack that weighed 15% BW, and a backpack that weighed 30% BW. A five-camera Vicon system was used in conjunction with two Kistler force plates (Goh et al., 1998). All body segments were assumed to be rigid and their movements were kinematically evaluated. The trunk flexion angle was measured based on the locations of the acromion process and anterior superior iliac spine. The matched kinematic and ground reaction force data was used in an inverse dynamics model to calculate the peak lumbosacral resultant force, compression force, and shear force during single stance phase. Each of the ten individuals performed at least three trials of each condition, and the most representative set of data was chosen for analysis based on 1) inter-trial gait consistencies 2) good force plate readings, and 3) clear marker positions (Goh et al., 1998).

Goh et al. (1998) concluded that every participant displayed compensatory trunk flexion, but there was no significant difference in the range of trunk motion between trials, averaging at approximately five degrees. There was no significant change in the walking speed or stride length with load. Carrying load disproportionally increased the forces at the lumbosacral joint; carrying a load that weighed 15% BW resulted in a 26% increase in the lumbosacral joint force and carrying a load that weighed 30 % BW resulted in a 64% increase in the lumbosacral joint force (Goh et al., 1998). This disproportionality is likely because of the increased trunk flexion due to added load, along with the added weight of the load. The mean values recorded for each variable measured by Goh et al. (1998), including walking speed, stride length, cadence, range of flexion-extension trunk motion, trunk flexion angle, and the peak lumbosacral joint resultant, compressive, and shear forces, are displayed in Table 2.3.

Table 2.3: Mean values compared between conditions from Goh et al. (1998)

Variable	No Load	15%BW Load	30%BW Load
Walking speed (m/s)	1.22±0.19	1.31±0.17	1.26±0.14
Stride Length (m)	1.27±0.18	1.43±0.10	1.25±0.35
Cadence (steps/min)	113±12.5	106±14.5	112±8.5
Range of Flex-Ext Trunk Motion (°)	4.72±0.93	5.10±1.37	5.51±0.97
¯Trunk Angle (°)	-8.38±1.40	-0.56±1.56	4.26±1.58
Abdominal Muscle Forces (% MVC) +	6±2	9±5	8±4
¯Peak L5-S1 Resultant Force (*BW)	1.50±0.24	1.90±0.27	2.46±0.29
¯Peak L5-S1 Compression Force (*BW)	1.48	1.87	2.42
¯Peak L5-S1 Shear Force (*BW)	0.28	0.35	0.45

¯Variables that demonstrated significant differences between each pair of load conditions
+ MVC: maximal voluntary contraction

IMPLICATIONS OF LOAD CARRIAGE

Through reviewing the literature on load carriage, it is evident that such an activity affects several anatomical structures and may pose a risk to the musculoskeletal system of the carrier. The bony skeleton is compressed, the muscles attached to the skeleton activate to control the load, and other soft tissue structures such as cartilaginous discs and ligaments help to transmit and support the load. All of these structures will adapt to the physical stress over time to become stronger and more resilient – unless the load is too great and causes a fatigue or acute injury first. Carriage of heavy loads implies high-to-severe conditions of locomotion (Beekley et al., 2007; Grenier et al., 2012; Knapik et al., 1996), and carrying a heavy backpack has been associated with several symptoms of discomfort. Muscle soreness, pain, numbness, shoulder pain, shoulder rounding, and spinal deformity, among other issues of the spine, have been linked to regular carriage of a heavy backpack (Atreya et al., 2010; Grimmer, Williams, & Gill, 1999; Negrini & Carbalona, 2002; Pratt, 1994; Sheir-Neiss, 2003).

Two suggested causes of symptoms of low back pain when carrying a load in a backpack are inadequate strength of the posterior trunk muscles and angular oscillations of the trunk (Vacheron, 1999). As expected, the majority of study participants ranked the backpack-free condition first for comfort, before three different backpack conditions (Gillet et al., 2006). These issues refer mostly to chronic carrying of a backpack, but more acute issues may also be a risk. Performing exercises with too much weight for the current level of strength of the individual can be dangerous and increase the risk of injury (Fahey, Insel, & Roth, 2007). Also, as mass is added posterior to the spine, the centre of gravity of the system shifts posteriorly away from its neutral position, and with greater displacement comes less stability (Birrell et al., 2007; Chow et al., 2005).

The positions of the load during carriage can significantly affect the mechanics behind the resulting joint movements and any resulting effects on the body. Carrying the entire load on the back involves different mechanics than dividing the load into front and back packs, for example. Compared to a purely posterior load, a more natural gait was adopted when a portion of the load was moved anterior to the trunk via three one kilogram sliding side pockets (Gillet et al., 2006). These packs were designed so the carrier could move some of the equipment to their front for easy access, and is biomechanically similar to carrying some of the load in a front pack (Gillet et al., 2006). In another study, a double-pack that carried equal loads anterior and posterior to the trunk was concluded to be more effective than a backpack (Kinoshita, 1985). As mentioned previously, carrying 16 kilograms in the webbing instead of the backpack lead to a greater vertical impact peak and a decrease in stance time (Birrell et al., 2007). This may be because the placement of some weight anteriorly likely balanced a portion of the posterior backpack weight, and moved the centre of gravity closer to neutral to allow for a more natural gait (Birrell et al., 2007). The frontpack–backpack concept was found to be most economical when the anterior and posterior loads were equal in weight (Legg, 1985). The results of these investigations of load distribution can be applied to expectations for the piggyback carry. Because a portion of the load, distributed in the arms and legs of the carried person, will be anterior to the trunk of the carrier, the biomechanical parameters measured during the piggyback may be more similar to a front-back pack than a backpack alone.

Compared to studies that have calculated lumbar spine forces in anterior-load lifting models or to kinematic studies that have evaluated the alterations in gait or trunk position with the addition of a posterior load, this study will address the trunk position during piggyback and backpack carriage and then estimate the lumbosacral forces present during these types of posterior load. If the trunk did not adjust into a flexed position to move the centre of gravity of the backpack or piggyback load forward to be within the base of support of the carrier, the load would create an extensor torque. Therefore, the piggyback is an interesting exercise that necessitates the coordination of the trunk flexors, which resist the extensor torque initially caused by the load to flex the trunk and pull the line of gravity anteriorly towards the base of support, and the trunk extensors, which prevent the trunk from collapsing into further flexion due to the force of gravity on the upper body once it has been moved forward.

This concept, which is a primary concern of the current study, is described by Figure 2 in the paper by Goh et al. (1998) and was redrawn and elaborated on as Figure 2.20. Figure 2.20 illustrates the presence and reason for a compensatory trunk flexion posture when carrying a load posterior to the trunk. This diagram illustrates how adding a loaded backpack moves the centre of gravity of the body posteriorly. In the following portion of the diagram, the trunk moves into a position of flexion to shift the centre of gravity of the body anteriorly back to its original position over the base of support at the feet (Goh et al., 1998). Based on this idea, it is thought that the current backpack condition will result in a position of greater trunk flexion than the current piggyback condition. This is because of the displacement of the shift in the location of the centre of gravity. Unloaded, the centre of gravity is near the anterior-posterior centre of the body within the lower torso (Goh et al., 1998). With the entire load posterior to the trunk, the backpack condition moves the centre of gravity of the backpack and person further posteriorly than does the piggyback condition, where some of the load was anterior to or in line with the original centre of gravity. Because the centre of gravity of the carried and carrier would be more anterior than the centre of gravity of the backpack and carrier, it does not require as much trunk flexion in order to shift the centre of gravity of the system back to the approximate location of the centre of gravity of the carrier when unloaded. This change of position is probably necessary in order to maintain balance and keep the centre of gravity within the base of support.

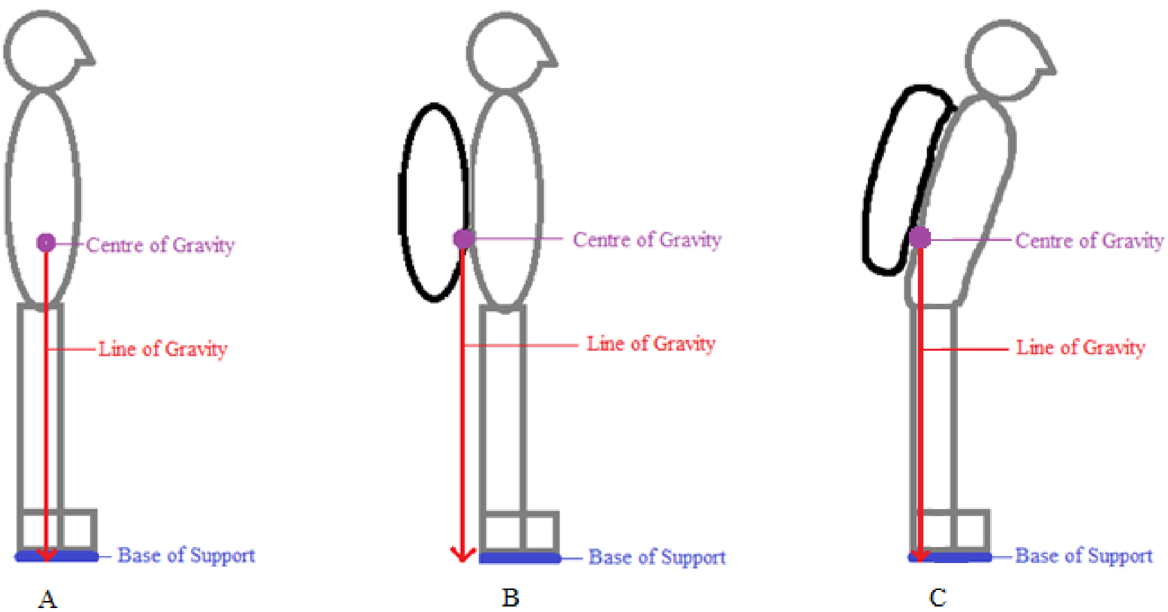


Figure 2.20: Trunk flexion is a response to the addition of a posterior load. A) The centre of gravity of the system is naturally located above the base of support. B) The addition of a posterior load shifts the centre of gravity of the system posteriorly, bringing the line of gravity potentially out of the base of support. C) The trunk flexes in order to shift the centre of gravity of the system forward and over the base of support. (Goh et al., 1998)

VARIABLES AFFECTED BY LOAD CARRIAGE

Because piggybacking has not been previously investigated in a scientific manner, the findings of backpack carriage will be the basis of the expectations for variables possibly affected by piggybacking. Based on these load carriage studies, it is evident that load can affect gait patterns, ground reaction forces, lumbar torque and forces. Gait parameters that have either sometimes or always been demonstrated to be affected by load are: walking speed, cadence, step length, stride length, the duration of the double and single support phases, and the duration of stance. The trunk flexion angle and range of motion should also be addressed, as well as the maximal hip and knee flexion. Important ground reaction force variables are the vertical ground reaction force maximal peaks and the dip between them. With respect to the lumbar forces and lumbar risk, researchers often address the lumbosacral compressive, shear, and resultant joint reaction forces. In the following paragraphs, each parameter will be outlined for its occurrence during normal gait and the common alterations seen during load carriage.

Gait Variables

The biomechanics of human gait have been well documented. One complete cycle of walking consists of several phases. During gait, single stance phase and double stance phase alternate, with each foot spending approximately 60% of a stride in stance phase and 40% of a stride in swing phase (Winter, 1991). Stance phase is composed of 15% heel strike, 15-30% flat-footed in mid-stance, and 30% propulsive phase (Winter, 1991). The greatest trunk flexion and greatest lumbar forces are exhibited near the end of single stance phase (Khoo et al., 1995), which will therefore be the moment of interest for much of the analysis. Gait is altered by the addition of a loaded backpack in several ways, likely decreasing gait efficiency to increase stability (Chow et al., 2005).

Walking speed, as seen in Equation 2 below, has sometimes been found to decrease while loaded (Chow et al., 2005). This was observed with loads of 10%, 15%, and 20% of body weight (Singh & Koh, 2009). Walking speed may decrease with added load as a mechanism to decrease the stress experienced by the lower limb (Birrell et al., 2007). Cadence is the number of steps taken in one minute, and a typical walking cadence is between 101 and 122 steps per minute (Winter, 1991). It has also been seen to decrease when a load is added, as demonstrated by Chow et al. (2005).

Along with walking velocity and cadence, step length and stride length have also been seen to decrease during load carriage (Chow et al., 2005; Atreya et al., 2010). Step length is

Equation 2: velocity (v) = $\frac{stride\ length * cadence}{120}$

where v represents the walking velocity in metres per second, stride length is the distance between successive stance periods in metres, and cadence is the number of steps per minute.

(Winter, 1991)

defined as the linear distance between successive points of floor to floor contact from one foot step to the following foot step by the opposite foot (Winter, 1991). Typically, step length decreases when carrying a load (Atreya et al., 2010; Chow et al., 2005). Stride length, or the linear distance between successive points of floor to floor contact of the same foot (Winter, 1991), has been measured by several researchers investigating the effects of load on gait (Myung & Smith, 1997; Cottalorda et al., 2003; Chow et al., 2005; Atreya et al., 2010). Normal unloaded stride length was 1.38m, 1.51m, and 1.64m for slow, average, and fast walkers respectively (Winter, 1991). The average stride length of male university students was approximately 1.48 m (Myung & Smith, 1997). Stride length may decrease when carrying a back pack, but no significant difference was found by Chow et al. (2005).

Load carriage increases the relative amount of time one spends in contact with the ground, measured by the durations of double support phase, single support phase, and or stance time. Double and single support phase durations are typically addressed as the percent of time spent in those phases with respect to the entire stride. The duration of double stance phase tends to increase with load (Birrell et al., 2007; Chow et al., 2005; Cottalorda et al., 2003). The duration of an unloaded typical double support phase is usually 8 to 11 percent of stride (Winter, 1991), but the double support duration increased from 14.2±1.1 to 15.6±1.1 percent of stride when loaded with only a ten kilogram backpack (Cottalorda et al., 2003). This is likely a protective mechanism to increase stability and decrease the risk of falling by decreasing the amount of time one leg only must support the body and load (Birrell et al., 2007). The single support time decreased (Chow et al., 2005) when carrying load, beginning at approximately 89 to 92 percent of stride when unloaded (Winter, 1991). Stance time is typically recorded in seconds, milliseconds, or as a percent of the stride period and has been seen to increase with load (Birrell et al., 2007; Cottalorda et al., 2003). It increased from 647±49 ms to 678±46 ms when carrying a ten kilogram backpack (Cottalorda et al., 2003). Stance time increased from carrying no load to carrying the load superior to the head, when the centre of gravity of the object was aligned with the centre of gravity of the body, to carrying a load posterior to the spine (Lloyd et al., 2011).

Kinematic Variables

Much research has been conducted on the effects of load carriage on gait kinematics. Some kinematic variables that may be useful for the present comparison between backpack and piggyback load are: the trunk range of motion, maximal trunk flexion angle, maximal hip flexion angle, and maximal knee flexion angle.

The Trunk

The trunk range of motion is minimal during gait. Several studies have found that there are only a few degrees of variation in the trunk angle throughout the duration of a normal gait cycle (Goh et al., 1998; Cappozzo, 1981; Thorstensson et al., 1984; Winter et al., 1993). The trunk range of motion averaged at five degrees when walking with no load, 15% BW, and 30% BW, regardless of the magnitude of loading (Goh et al., 1998).

Although usually approximately upright when walking, the trunk and neck have been seen to increase in flexion and decrease in lumbar lordosis when carrying a substantially loaded backpack (Hall, 2007). The estimated angles of trunk flexion vary by study, which is assumed to be due to different methods of measurement, error or individual variation, but the comparisons from unloaded to loaded all show an increase in trunk flexion with load. Without any load, trunk flexion angle was approximately 4.9 +/- 2.7 degrees during gait, but with a 15 kg backpack, the trunk flexion angle was approximately 14.2 +/-3.2 degrees (Gillet el al., 2006). Gillet et al. (2006) suggested that an increase of six to eleven degrees of trunk flexion compensates for the backwards rotating moment of a posterior backpack load. When part of the load is carried anterior to the trunk, trunk flexion may be approximately 10.6 +/- 2.9 degrees, as was measured when there was 12 kg posteriorly and 3 kg anteriorly (Gillet et al., 2006). The average trunk angles exhibited while walking with no load, a 15% BW backpack, and a 30% BW backpack were 8.38±1.4 degrees of extension, 0.57±1.56 degrees of extension and 4.26±1.58 degrees of flexion respectively (Goh et al, 1998). Kinoshita (1985) and Rodriquez-Soto et al. (2013) also documented increased trunk flexion while walking with both relatively light and heavy loads, and the lumbar spine was specifically seen to flex during loading via magnetic resonance imaging (Rodriquez-Soto et al., 2013). However, these effects may be able to be reduced by keeping the centre of gravity of the load low in the backpack and by carrying no greater than 15% BW (Hall, 2007).

The Hip and Knee

The results of a few studies suggest that there is a decrease in transverse plane motion and an increase in sagittal plane motion in the lower half of the body when walking with a load. This has mostly been documented for the hip, but an increase in sagittal plane motion will likely also affect peak knee flexion so it will also be considered a variable of interest. Adding weight to a backpack increases the moment of inertia of the backpack, which increases the impulse necessary to initiate motion as well as the moments necessary to control motion (Atreya et al., 2010). Because the weight of the bag will need to be controlled, the movement patterns will be altered to make this an easier task. This increase in sagittal plane motion is logical. Upon every heel strike, the mass of the system that was moving inferiorly is resisted by the ground, stopped, and begins to move superiorly during the propulsion phase. This sudden impact transmits a force through the body. When a load is added, this impact peak increases and there is an increased demand for shock absorption (Chow et al., 2005). The impact exerts an impulse, or a force over

Equation 3: Impulse = Force * time

*where the impulse is measured in N*s, force is measured in N, and time is measured in s.*

(Hall, 2007)

a time as seen in Equation 3, on the body. In order to decrease the force component of the impulse, the duration over which the force is exerted can be increased. Therefore, it is logical that the carrier may instinctively allow more flexion at the hip or the knee, which would allow for a greater time to absorb the impact and would decrease the force experienced. This potential transition from more transverse to more sagittal plane motion is likely to increase stability. Because an increase in sagittal plane motion may be expected, maximal hip flexion and knee flexion angles may increase with load (Birrell et al., 2007; Chow et al., 2005). Chow et al. (2005) recorded significant increases in the peak knee flexion during the loading response and in peak hip flexion and peak hip extension during stance, when carrying a backpack weighing 15% BW. When an increase in these sagittal plane movements occurs, the vertical ground reaction force pattern would also likely be affected.

Kinetic Variables

Ground Reaction Forces

Ground reaction force patterns are cyclic, repeating every gait cycle, and can be measured during stance by a force platform. A ground reaction force is the force exerted by the ground on the foot in the opposite direction to an equal force of the foot pushing on the ground (Hall, 2007). Ground reaction forces may be exerted in the vertical, anteroposterior, or mediolateral directions, or a combination of these directions (Robertson et al., 2004). Because this paper is investigating biomechanics in the sagittal plane and vertical motion may alter with load, only the vertical ground reaction forces will be discussed.

The bimodal vertical ground reaction force can be illustrated by a force-time curve, as seen in Figures 2.21 and 2.22. If one exists, the first spike in vertical ground reaction force corresponds to heel strike, which is illustrated in Figure 2.22 but not Figure 2.21. Next, the force increases to the first maximal value that represents weight acceptance in general, known here as the first peak (Robertson et al., 2004; Winter, 1991). This force decreases into a dip that represents the support flexion phase before increasing to its second peak, which represents the propulsive phase (Robertson et al., 2004; Winter, 1991).

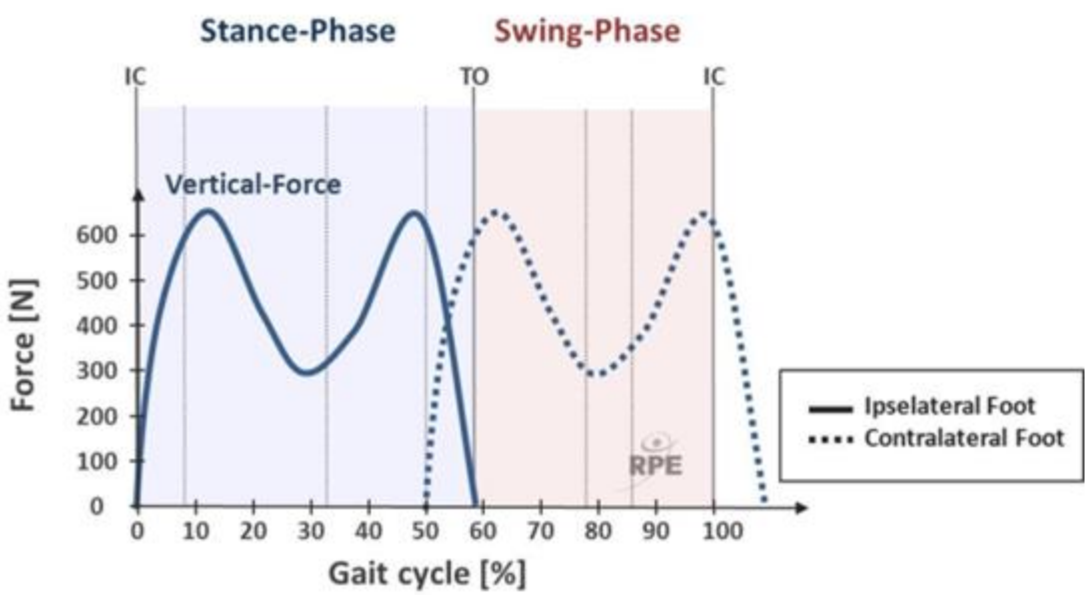


Figure 2.21: A vertical ground reaction force curve, as measured by a force plate. IC represents initial contact and TO represents toe-off. (Jakarandatree, 2012)

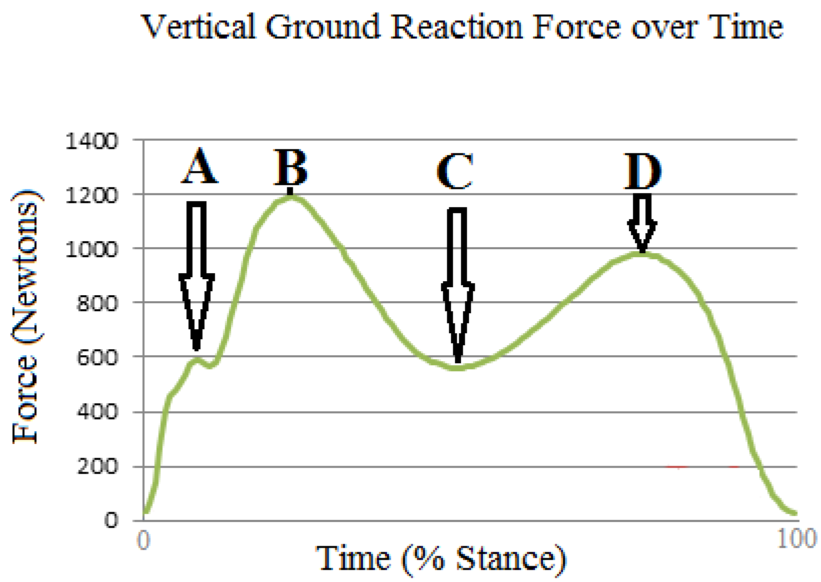


Figure 2.22: A vertical ground reaction force curve from an unloaded trial during the pilot study. A) initial impact peak, B) 1st maximal peak at weight acceptance, C) dip between peaks during support flexion, D) 2nd maximal peak at propulsion.

Because mass varies by participant, the raw graphs of vertical ground reaction force in Newtons over time may not be directly comparable between participants. However, division normalization helps solve this problem (Wannop et al., 2012); each value on the y axis is divided by the mass of the participant to create a new y value. These new y values are then plotted in terms of body mass ($N*BM^{-1}$) as some studies have done (Birrell et al., 2007; Chow et al., 2005), such as 1.5 times body mass for example. Therefore, the vertical ground reaction force (vGRF) may be described in terms of force in Newtons, normalized to Newtons per kilogram (Winter, 1991) or compared to body weight (Robertson et al., 2004), the latter two of which enable vGRF to be compared across participants. Because this is a manipulation of the data and may introduce error, normalization should only occur when necessary and not when comparing within participant data (Wannop et al., 2012).

A few examples of values seen in a graph of vertical ground reaction force are to follow. Although the weight of the participant was not noted, an example of possible force values was illustrated by Robertson et al. (2004) and will be described. The ground reaction force begins at near zero and increases sharply to a small peak of 790 N at approximately 0.05 seconds and a maximum of 900N at approximately 0.12 seconds. There is then a gradual decrease to 430 N halfway through stance and a mirrored increase to approximately 800 N 0.12 seconds from toe-off (Robertson et al., 2004, 94). In Newtons per kilogram during a slow walk, the first peak may reach approximately 9.9, decrease to 8.5, and increase to 10.2 for the second peak (Winter, 1991). Average values for these points during an average cadence were 10.7, 7, 10.7 in Newtons per kilogram (Winter, 1991). These maximal values generally increased during a brisk walk, with these points altering to approximately 12.5, 5.5, 11.8 (Winter, 1991). Vertical forces hover around body weight, fluctuating by approximately 30% of body weight (Robertson et al., 2004). Example values using force (N) per body weight (N) were approximately 1.1 at the first peak, decreased to just less than 0.8 to 0.85 between the peaks, and increased again to 1.06 to 1.2 at the second peak (Robertson et al., 2004; Scott-Pandorf, Stergiou, Johanning, Robinson, Lynch, & Pipinos, 2007).

Ground Reaction Forces during Load Carriage

With the addition of load carriage, the vertical ground reaction forces increase. Several studies found that the first peak increased (Cottalorda et al., 2003; Gillet et al., 2006; Chow et al., 2005), the second peak increased (Cottalorda et al., 2003; Gillet et al., 2006; Chow et al., 2005), and the average force increased (Cottalorda et al., 2003) when a load was added posterior to the spine. Chow et al. claim that the increase in vertical ground reaction force is only because of the load and not because of any gait changes (2005). It is logical to expect that an increase in weight would cause a proportional increase in the force supporting the weight. For every kilogram of added load, there was an increase of almost ten Newtons of vertical ground reaction force (Birrell et al., 2007)(Lloyd et al., 2011). This corresponds to Equation 4 that states that force equals the product of mass and acceleration, with an increase of force of approximately ten

Equation 4: Force (F) = Mass (m) *Acceleration (a)

where force is measured in N, mass is measured in kg, and acceleration is measured in m/s^2 (Robertson et al., 2004)

Newtons when an added mass equals one kilogram and acceleration equals 9.81 meters per second squared (Robertson et al., 2004). Birrell et al. (2007) agreed that vertical ground reaction forces did seem to increase proportionally when eight kilogram increments of weight were added to a loaded backpack (Birrell et al., 2007). Although these researchers did not find significant results with respect to changes in vGRF with load, the pilot study participants felt quite different and more comfortable carrying a person versus a backpack, and it is thought that they may have altered in gait. An alteration in gait, whether it be to decrease movement in the transverse plane and increase movement in the sagittal plane (Birrel et al., 2007) or to walk with a stiffer body and greater force transmission when load is added (Holt et al., 2003; Tilbury-Davis & Hooper, 1999), should theoretically alter the ground reaction force curve. Also, it appeared that the initial impact peak was only present during unloaded trials during the pilot study in June 2013.

Ground Reaction Force Data

Force plates are a sophisticated tool commonly used in Biomechanics research, often as part of gait analysis or spinal force estimations. Force plates are usually built into the walkway, as are the force plates at Pan Am Clinic Foundation seen in Figure 2.23. When the sensors within the force plate are compressed, the force plate takes readings of the magnitude and direction of the ground reaction forces present and these readings can be graphed over time. There are several types of force plates but two common brands are Kistler (Birrell et al., 2007; Gillet et al., 2006; Goh et al., 1998; Kinoshita, 1985; Khoo et al., 1995; Wannop et al., 2012;) and Advanced Mechanical Technology, Inc (AMTI) (Chow et al., 2005). Force plate data can be recorded at different frequencies. Different researchers have used frequencies ranging from 60 Hz (Chow et al., 2005) to 100 Hz (Atreya et al., 2010) to 400 Hz (Birrell et al., 2007; Kinoshita, 1985) to 2400 Hz (Wannop et al., 2012), depending on the sophistication of the system and use of the data. Even while using a six-camera Vicon system, Chow et al. recorded and synchronized video and force plate data at 60 Hz (2005). In general, 60 Hz has been found to be acceptable for gait analysis data collection as stated by Myung and Smith (1997).

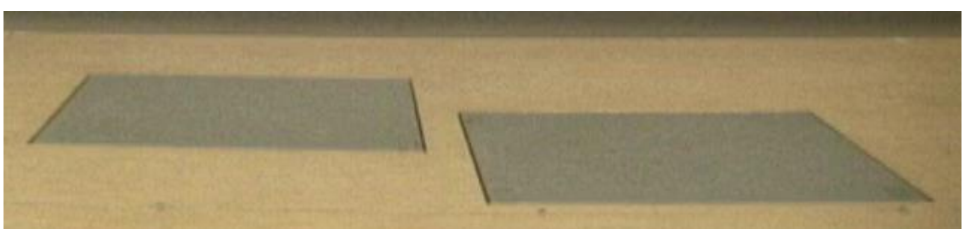


Figure 2.23: Two AMTI force plates embedded in the floor of the Pan Am Clinic Foundation Biomechanics Laboratory

Weight of the Load

In the military, the weight of a carried load often exceeds 50% of lean body mass (Beekley, 2007) and can regularly reach 60% of body weight (Birrell et al., 2007) or 68 kilograms (Rodriguez-Soto, 2013), but this magnitude of load has rarely been studied biomechanically (Beekley, 2007). Several previous studies that involved load carriage used maximum loads ranging from 21.4 kilograms to 43 kilograms (Epstein et al., 1987; Epstein et al., 1988; Vacheron, 1999; Birrell et al., 2007; Chatterjee et al., 2012; Grenier et al., 2012;

Rodriquez-Soto, 2013), or 40 % of body weight (Myung & Smith, 1997), or 70 % of lean body mass (Beekley, 2007). One study that used a maximum of 40 kilograms ensured that their rear-foot striking participants had a mass greater than 70 kilograms, had experience carrying loaded backpacks, and had passed a health screen questionnaire before they could be included (Birrell et al., 2007).

Some load carriage studies (Epstein, 1988; Myung & Smith, 1997; Simpson, Munro & Steele, 2011; Grenier et al., 2012) had participants performing an activity of greater risk while carrying the load rather than simply walking for several metres. Epstein et al. (1988) had participants walk for 120 minutes on a treadmill with 40 kilograms of load. Myung & Smith (1997) had participants in harnesses walking on a slippery floor while carrying a 24 kilogram box. In another study, fifteen female hikers carry loads weighing 0%, 20%, 30%, and 40% of body weight over an eight kilometer distance (Simpson, Munro, & Steele, 2011). In 2012, a group of researchers had participants carry up to 43 kilograms for 21 hours of military simulation in mountainous terrain (Grenier et al.). Potvin et al. (1991) based their loads on the maximum permissible limit NIOSH guideline of 6400 N of disc compression forces.

It is evident that no specific load limit has been agreed upon, but limiting the mass of hiking packs has been provided as a strategy to prevent hiking injuries. It is difficult to determine a limit that would apply to all contexts. Recreational hikers may carry their packs for long durations over a shorter amount of days and military personnel may have to carry their packs for long durations over many days. More specifically, after studying heavy loads carried by female hikers, a 30% of body weight limit was suggested to restrict the risk of injury to the lower limbs (Simpson et al., 2011). Cathcart et al. (1923) recommended a maximum carrying weight of 40% of body weight (as cited in Myung & Smith, 1997), and Snook (1978) recommended 24 kg as a maximum acceptable weight of carry for 90% of male industrial workers in 1978 (as cited in Myung & Smith, 1997). Although limits may be suggested, hiking and military situations are usually such that the load tends to be whatever needs to be carried. However, athletes and their athletic trainers are often able to choose the weight that they carry for training purposes, depending on their sport, and it can be decreased if necessary. For example, a lighter person may be able to be carried during piggybacking drills. The most valid guide for a safe magnitude of weight load is starting with something that can be lifted for at least eight repetitions and then a logical progression (ACSM, 2013; Fahey, Insel, & Roth, 2007).

Lumbosacral Loads

As previously mentioned, the magnitude of the lumbosacral forces are affected by the position and amount of load carried (Cholewicki et al., 1991). When weight is carried on the back, the trunk tends to increase in flexion in order to shift the centre of gravity anteriorly so that it is over the base of support (McGill, Marshall, & Anderson, 2013). When the position of the body is altered, the location of the line of gravity shifts and the forces on the lumbar spine change accordingly (Levangie & Norkin, 2005). A less upright torso increases the moment arm of the weight, which then increases the forward torque caused by the load and necessitates an increase in the magnitude of the erector spinae muscle force (Cholewicki et al., 1991). Both the increase in load and the increase in muscle force will increase the joint reaction forces estimated for the lumbosacral joint. There are two primary methods that can be used to determine the loads on the spine: direct measurement using invasive methods such as a needle to measure the intradiscal pressure, and indirect calculation using either electromyography, biomechanical modelling, or a combination of both (Khoo et al., 1995).

Biomechanical Lumbar Back Model – The Lumbosacral Joint

Considered a reflection of the stress experienced by the lumbar spine in a certain position or while sustaining a specific load, the compressive, shear, and resultant joint reaction forces are often estimated using a lumbar spine model (Nordin & Frankel, 2012). These models are based on the anatomy of the lumbar spine and certain assumptions need to be adopted, such as the length of the moment arm or angle of pull of the extensor muscles (Bogduk et al., 1992; McGill & Norman, 1987). The lumbosacral joint is often selected as the joint of interest and the trunk flexion-extension axis of rotation in lumbar spine lifting models (Hall, 2007; Khoo et al., 1995), usually because it is comparatively vulnerable as noted previously. Biomechanical models are either dynamic, used when segment accelerations are deemed important, or static, used when there are little to no accelerations present (Chaffin et al., 1999).

Static models are based on the concept that zero acceleration allows for equilibrium equations to be utilized. With zero acceleration of the system, the sum of the torques is set to zero and the torques are balanced to solve for the unknown, which is the resultant muscle torque in this case. This resultant torque is then converted to resultant force using the torque equation seen in Equation 1, and the sum of the forces in each axis are set to zero. Assuming that all of the forces in the y axis sum to zero and that all of the forces in the x axis sum to zero allows the compressive joint reaction force, in the y axis, and the shear joint reaction force, in the x axis, to be calculated. The Pythagorean Theorem may then be used to translate these forces to the resultant joint reaction force (Nordin & Frankel, 2012); all of these forces are often compared between activities to evaluate relative lumbosacral joint risk (Kavcic et al., 2004; Cholewicki et al., 1991).

Several researchers have used static models. Cholewicki et al. (1991) used a static analysis while estimating the L4-L5 moments and forces during the powerlift deadlift, McLaughlin et al. (1978) found a difference of less than ten percent when comparing static and dynamic estimations for a different powerlift, and Lander et al. (1990) estimated a 1% difference between using quasi-static and dynamic models during a squat analysis. Static models have frequently been used to address the kinetic effects of lifting on the lumbar spine (McGill & Norman, 1985). Some other sources that have used static lumbar models are Bejjani et al. (1984), Chaffin et al. (1999), Cholewicki et al., (1991), Enoka (1979), Roozbazar (1975), and Schultz et al. (1982). While walking with a loaded backpack, the average range of motion of trunk flexion-extension is five degrees (Goh et al., 1998), which is minimal considering the potential for measuring error. Because the accelerations are minimal, the present study simplified the upper body as demonstrating an approximately constant position of trunk flexion and assumed equilibrium.

There are several common techniques for determining the centre of gravity of an object, including the reaction board technique, the suspension method, balancing on a knife-edge, and a mathematical segmental method (Ozkaya, Nordin & Leger, 2012). Because of the availability of the equipment needed to perform the reaction board technique and the logistics of other techniques, the reaction board technique was used to determine the location of the centre of gravity of the backpack.

This technique is based on the zero acceleration of a static system, and involves balancing torques. The reaction board method requires a horizontal board with a knife-edge on the bottom side of both ends, one end resting on a platform and the other resting on a scale as seen in Figure 2.24. Any weight read by the scale is said to be a partial weight because part of the weight is also supported by the platform. The partial weight of the board is known, as seen on the scale, and the distance between the knife edges is known, as per direct measurement. The knife-edge that is supported by the platform is considered the axis of rotation and the knife-edge on the scale will be two metres from it. A bag can be placed so one end is at the axis of rotation, and the partial weight of the bag plus the board can be recorded, then the partial weight of the board itself can be subtracted to leave the partial weight of the bag. Essentially, the torque produced by the partial weight of the bag at two metres (F_{Gp} BP * 2) can be set equal to the known weight of the bag at its unknown distance from the axis of rotation (F_G BP * d⊥BP); the unknown distance is then solved. To complete one measurement, this procedure is repeated two times, each with a new orientation of the bag, in order to locate its centre of gravity in the anteroposterior and superior-inferior dimensions.

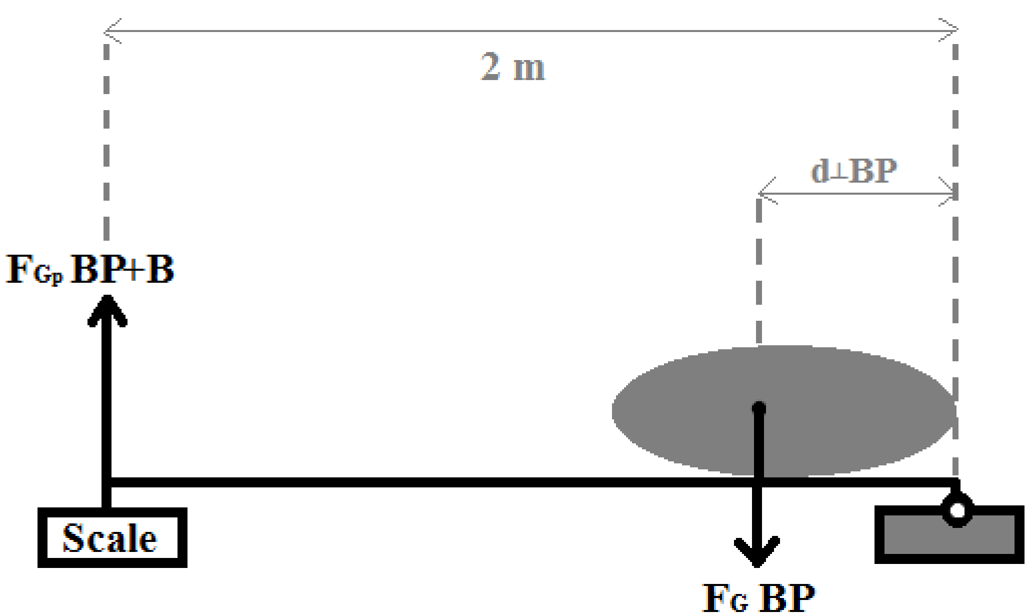


Figure 2.24: The Reaction Board Technique for Determining the Location of the Centre of Gravity. FGp BP+B represents the partial weight of the backpack and the board. F_GBP represents the weight of the backpack. The axis of rotation is represented by the white circle outlined in black. d⊥ BP represents the moment arm of the centre of gravity of the backpack, and the 2m distance represents the moment arm of FGp BP+B. Because $\Sigma T = 0$,
(FGp BP+B)(2)/(F_GBP) = d⊥ BP.

CHAPTER III

METHODS

DESCRIPTION OF STUDY

Participants

Twelve fit adult males with an average mass of 82.4±8.3kg volunteered as load carrier participants. Calculated for the use of an ANOVA with an alpha value of 0.05, an effect size of 0.4, three groups, three measurements, and a 0.5 correlation among repeated measures, G*power software determined that twelve participants were necessary for a power level of 0.80 (Faul, Erdfelder, Buchner, & Lang, 2009). The results of a similar study, which compared trunk flexion between walking with a posterior (15 kg) backpack load and walking with a mostly posterior (12 kg) but partly anterior (3 kg) backpack load, were considered when making this calculation (Gillet et al., 2006). Gillet et al. (2006) measured trunk flexion angles of 14.2±3.2° for the posterior backpack load and 10.6±2.9° for the partly anterior load.

All participants were resistance-trained fit adult males, and nine of the twelve were CrossFit athletes. Local CrossFit facilities were located via the internet before their offices were contacted by email and in person. CrossFit Winnipeg and CrossFit84 allowed the researcher to hand out a recruitment bulletin and talk to a couple of classes at which athlete email addresses were collected. The recruitment bulletin was also posted on facebook and all interested volunteers were emailed with additional information. Recruitment was aimed at CrossFit males who had used the piggyback carry, but the criteria for inclusion was: a male athlete with a body mass of at least 70 kg, regular physical training that involved lifting or carrying a minimum of 40 kg at least once a week, and no musculoskeletal injury within the last six months.

A 29 kilogram female child, who was located through family friends, was used as the passenger load for the piggyback condition.

Weight of the Load

One goal of the present study was to use the heaviest load that was not considered a high risk to the carriers. Based on the more difficult load-carrying tasks and much heavier loads carried in related studies (Beekley, 2007; Birrell et al., 2007; Grenier et al., 2012; Myung & Smith, 1997; Vacheron et al., 1999), it was considered reasonable to ask these weight-trained participants to walk less than ten meters several times while carrying twenty nine kilograms of load. Dependent on condition, this load was in the form of softner salt packages and weight plates in a hiking backpack, or a child on the back in the position of a piggyback. Because of the fitness and strength levels of these athletes and the relatively low load that was used in this study, the risk was seen as minimal when compared to similar studies (Birrell et al., 2007; Goh et al., 1998).

TESTING PROTOCOL

An Overview

Testing occurred at the Pan Am Clinic Foundation Biomechanics Laboratory in one session on April 8, 2014. The passenger was asked to come first and participants were scheduled to arrive in 15 minute planned intervals. First, each participant read information about the study and had the opportunity to ask questions. They confirmed that they fit the inclusion criteria, signed the informed consent form, and the carriers filled out the Participant Form. Carriers received one of twelve unlabelled envelopes, half of which stated “NL – BP – PB” and half of which stated “NL – PB – BP”. This randomly determined the order of the load conditions. Markers were attached to various locations on the carrier’s body, as described later. Height and body mass were measured on a scale and recorded. After the child passenger was weighed, the bag weight was fine-tuned accordingly.

The participants walked unloaded a few times to become familiar with the experimental set-up and to prepare the body for activity. During testing, each participant performed three successful trials of the unloaded condition first, so as to continue the familiarization and warm-up, before performing three trials for each of the two equally-weighted loaded conditions. The participants were allowed several practice trials per condition if they needed to become accustomed to the new load condition with the location of the force plate. The passenger tried to maintain a static position throughout each piggyback trial.

During each trial, the participant walked naturally across the eight meter walkway with the embedded force plate, while being video-recorded by cameras on both the right and left sides. This set-up can be seen in Figure 3.1 below. A trial was considered successful if the gait was consistent and natural (Birrell et al., 2006) with the right foot contacting the force plate, and only the force plate (Chow et al., 2005; Birrell et al., 2006; Wannop et al., 2012), during the step in question. Each participant performed at least three successful trials of every load condition before the next participant began trials.

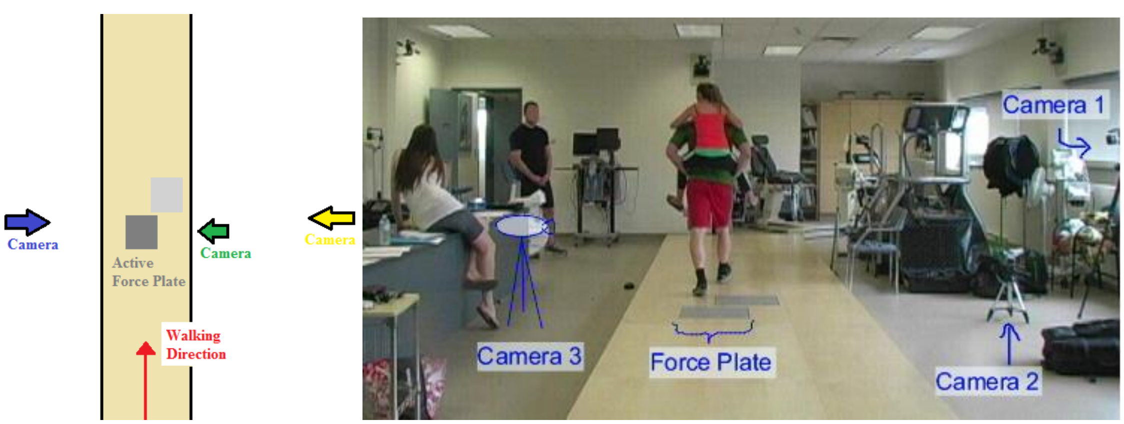


Figure 3.1: Experimental Set-up

Marker Placement

Green markers were placed on bony landmarks and beige markers were placed on the centres of gravity of two body segments, as seen in Figure 3.2 but on both sides of the body. The markers were one inch square pieces of painters' tape or masking tape and the appropriate locations were identified by trained research assistants. Common bony landmarks were identified by palpation and then marked with green tape, including the acromion process of the scapula, the anterior superior iliac spine, greater trochanter of the femur, lateral epicondyle of the femur and lateral malleolus of the fibula.

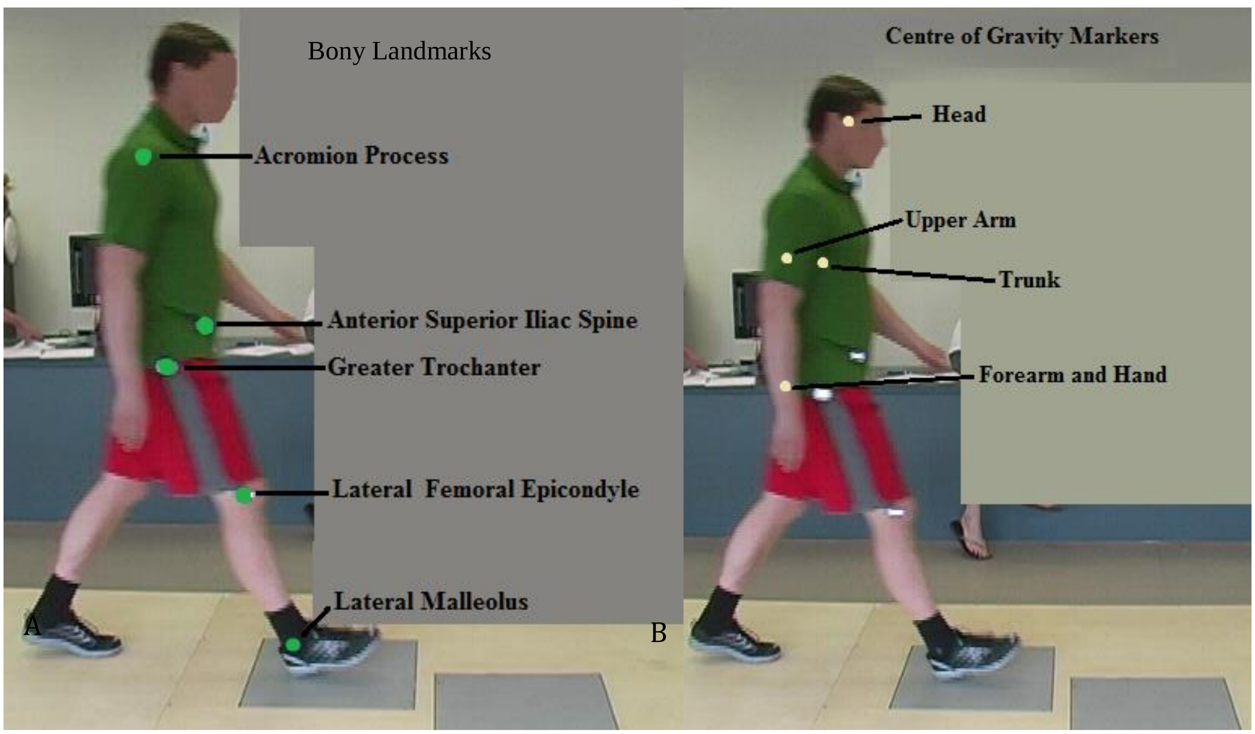


Figure 3.2: Position Markers. A) Green tape will identify the bony landmarks and B) Beige tape will identify the location of the centre of gravity of important limb segments.

The centre of gravity was only physically marked on the upper arm and the forearm and hand, while the trunk and head were also marked digitally later on the Dartfish software. In order to identify the location of the centre of gravity of each body segment, the segment endpoints were palpated, the distance between the endpoints was measured, that length was multiplied by the corresponding percentage found in Table 3.1, and the resulting distance was measured and marked from the proximal endpoint. The upper arm was measured between the acromion process and the head of the radius, and the forearm and hand, in accordance with the percentage, was measured from the head of the radius to the styloid process of the radius. Digitally, the head was measured from the trunk-neck intersection to the top of the head and the trunk was measured from the trunk-neck intersection to the greater trochanter. Only the segments listed above were marked on the carrier; additional segments were also marked on the passenger during digital analysis because their masses were superior to L5/S1 and therefore also influenced the torque experienced by the lumbosacral joint. These additional segments were the thigh from the greater trochanter to the knee joint line, the lower leg from the knee joint line to the lateral malleolus, and the foot from the heel to the tip of the great toe.

Table 3.1: Location of the centre of gravity and the relative weight of each relevant body segment by sex – Rearranged from Hall (2012) and Robertson et al. (2004)

Segment	% Length from Proximal Joint +		% Total Body Weight *
	Male	Female	
Head	55.0	55.0	7.28
Trunk	63.0	56.9	50.70
Upper Arm	43.6	45.8	2.63
Forearm & Hand	(62.58)*	(62.58)*	2.27
Thigh	43.3	42.8	10.27
Leg	43.4	41.9	4.35
Foot	50.0	50.0	1.47

+ Hall (2012)
*Robertson et al. (2004)

Determining the Location of the Centre of Gravity – The Backpack

After testing, the backpack was taken to the Biomechanics Laboratory at the University of Manitoba so the location of its centre of gravity could be determined using the reaction board technique. The backpack was placed on the scale and mass was confirmed. Next, the scale was placed under one side of the board while textbooks were placed under the other side to ensure that the board was horizontal. The partial weight of the board was recorded. The backpack was placed on the board and the partial weight of the board plus the backpack was measured. Refer to page 91 of Chapter II for details about the calculations; the distance from the axis of rotation at the end of the board to the centre of gravity of the backpack was solved and a string was held up for the Figure 3.3 photo to document the location of the centre of gravity in the transverse

Figure 3.3: Using the reaction board technique to locate the centre of gravity of the backpack in the transverse plane.

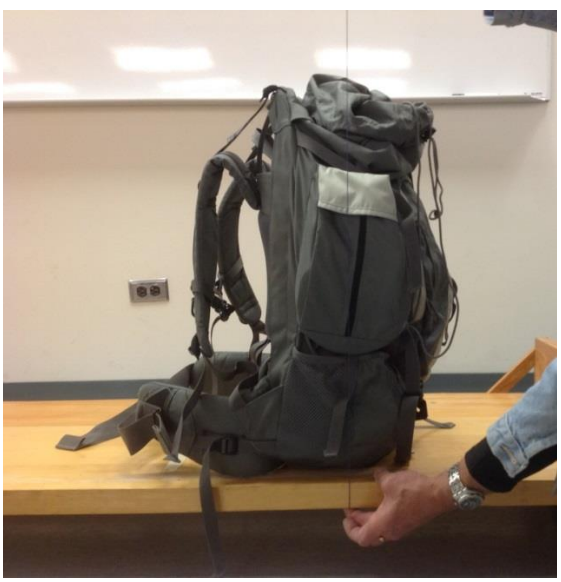


Figure 3.4: Using the reaction board technique to locate the centre of gravity of the backpack in the frontal plane.

plane. The backpack was then rotated, the new scale reading recorded, and calculations solved for the distance from the axis of rotation to the centre of gravity of the backpack in the frontal plane. A string was held up for the Figure 3.4 photo. The photos were superimposed to identify the exact location of the centre of gravity in two dimensions or the sagittal plane, which was necessary for later video analysis of the pack being carried.

Data Collection – Force Plate and Filming

A force plate was used in conjunction with three video cameras and motion analysis software to measure kinematic, gait, and ground reaction force parameters. The embedded Advanced Mechanical Technology, Inc (AMTI) force plate measured the ground reaction forces in three dimensions over time at a frequency of 200 Hertz; only the vertical dimension data was used during analysis. The data was transferred to the DELL Latitude E6540 laptop research computer in the form of Excel spreadsheets.

Three cameras filmed the participants at 30 frames per second: two Canon D2L cameras approximately 2.5 meters to either side of the individual as permitted by the laboratory space and one Canon HDV 1080i camera one meter away from the right side of the force plate for a closer view of the feet. This zoomed-in view was only meant to be used in the case of bad video from the other cameras, but was not needed. The right lateral view was used for all kinematic analysis except for any restrictions, such as the location of the centre of gravity of the left arm when it was hidden behind the trunk. The left lateral view camera was used for such occurrences. All video data was uploaded to the Dell research computer and analyzed using 2014 Dartfish Software 6 TeamPro 6.0 (Dartfish, 2014).

Data Analysis – Variables Measured

Kinematic, lumbosacral load, vertical ground reaction force and gait variables were measured for each trial. Each variable was measured at the position determined to be most relevant to biomechanical risk for that specific variable, which is described for each below. All variables were expected to provide insight into the sagittal plane motion that occurs when carrying piggyback and backpack loads. For each participant, results from the three successful trials of one condition were averaged to represent that condition (Chow et al., 2005).

Kinematic Variables

A key variable measured in this study was maximum trunk flexion angle, but minimum trunk flexion angle and the trunk range of motion were also measured. The maximum trunk flexion angle affects the moments experienced by the lumbosacral spine and is therefore essential to the purpose of the study and relevant with respect to back health. The instant of maximum trunk flexion was identified by scrolling frame by frame and marked as a key position in the right lateral view video in Dartfish. As seen in Figure 3.5, trunk flexion angle was measured between the vertical axis and the line of the trunk segment using the angle measuring tool in Dartfish. As the video was investigated frame by frame, the moment of least trunk flexion (or maximum extension) was also made a key position and measured in the same way. The difference between these two positions was calculated to determine the range of trunk motion. Trunk range of motion was an important variable; the lumbar spine model uses equilibrium equations that assume that the trunk has a minimal range of motion during gait. The maximal trunk flexion angle for each trial was also used in the lumbar load calculation for that trial.

Figure 3.5: Illustration of trunk, hip, and knee flexion angles.

Because a possible increase in sagittal plane motion with load was expected (Birrell et al., 2005; Chow et al., 2005), maximal hip flexion and knee flexion angles were also measured. The maximum hip flexion and maximum knee flexion angles seen in Figure 3.5 were identified by moving frame by frame, and were quantified using the angle measuring tool in Dartfish. The hip flexion angle was measured between the line of the thigh and the imaginary line extended inferiorly from the line representing the trunk. The knee flexion angle was measured between the line of the lower leg and the imaginary inferior extension of the line of the thigh. All angles were measured in degrees.

Kinetic Variables – Lumbosacral Load Variables

The position of greatest trunk flexion was assumed to be the position of greatest stress on the lumbosacral spine. Therefore, lumbosacral loads were calculated for the instant of maximal trunk flexion using a static biomechanical lumbar spine model. The magnitudes and directions of the unknown resultant muscle moment, the resultant muscle force, and the compressive, shear, and resultant joint reaction forces at the lumbosacral joint were calculated using video position data (Chaffin et al., 1999; McGill et al., 2009a, McGill et al., 2009b), anthropometrics, and mathematical calculations (Chaffin et al., 1999). The key position of maximum trunk flexion, which was previously identified in the right lateral view video in Dartfish, was used for all lumbar load analysis except for when left body segments were obscured by the trunk, passenger, or backpack, in which case the left lateral view video was used. Anthropometric data was used to determine the locations of the centres of gravity of each body segment and the weights of each body segment. A formula was created in Microsoft Excel and was used to calculate the lumbar load variables.

The Lumbosacral Spine Model

The location of the lumbosacral joint was estimated and marked on-screen in Dartfish based on previous research and images of the external view of the body (Moore et al., 2010). The anterior-posterior width of the carrier was measured superior to the iliac crest, slightly superior to the location that would be imagined to present a skin dimple over the posterior

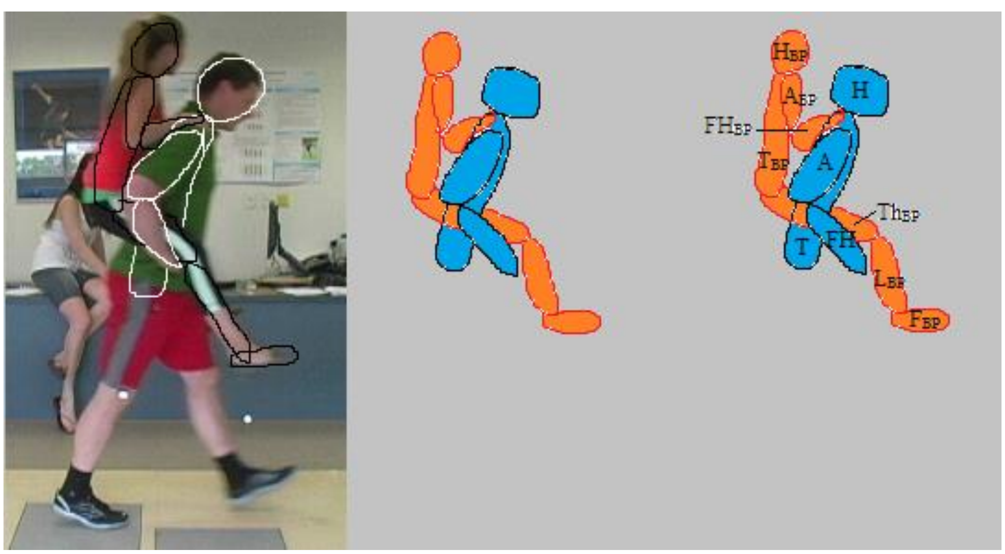


Figure 3.6: The body segments used for piggyback analysis: the head (H), trunk (T), upper arm (A), and forearm and hand (FH) of the carrier, as well as the head (H_{BP}), trunk (T_{BP}), upper arm (A_{BP}), forearm and hand (FH_{BP}), thigh (Th_{BP}), lower leg (L_{BP}), and foot (F_{BP}) of the person being carried.

superior iliac spine (Khoo et al., 1995). The joint was marked at one third of this anterior-posterior distance from the posterior surface of the trunk; this marker represented the mediolateral axis of rotation of the lumbosacral joint.

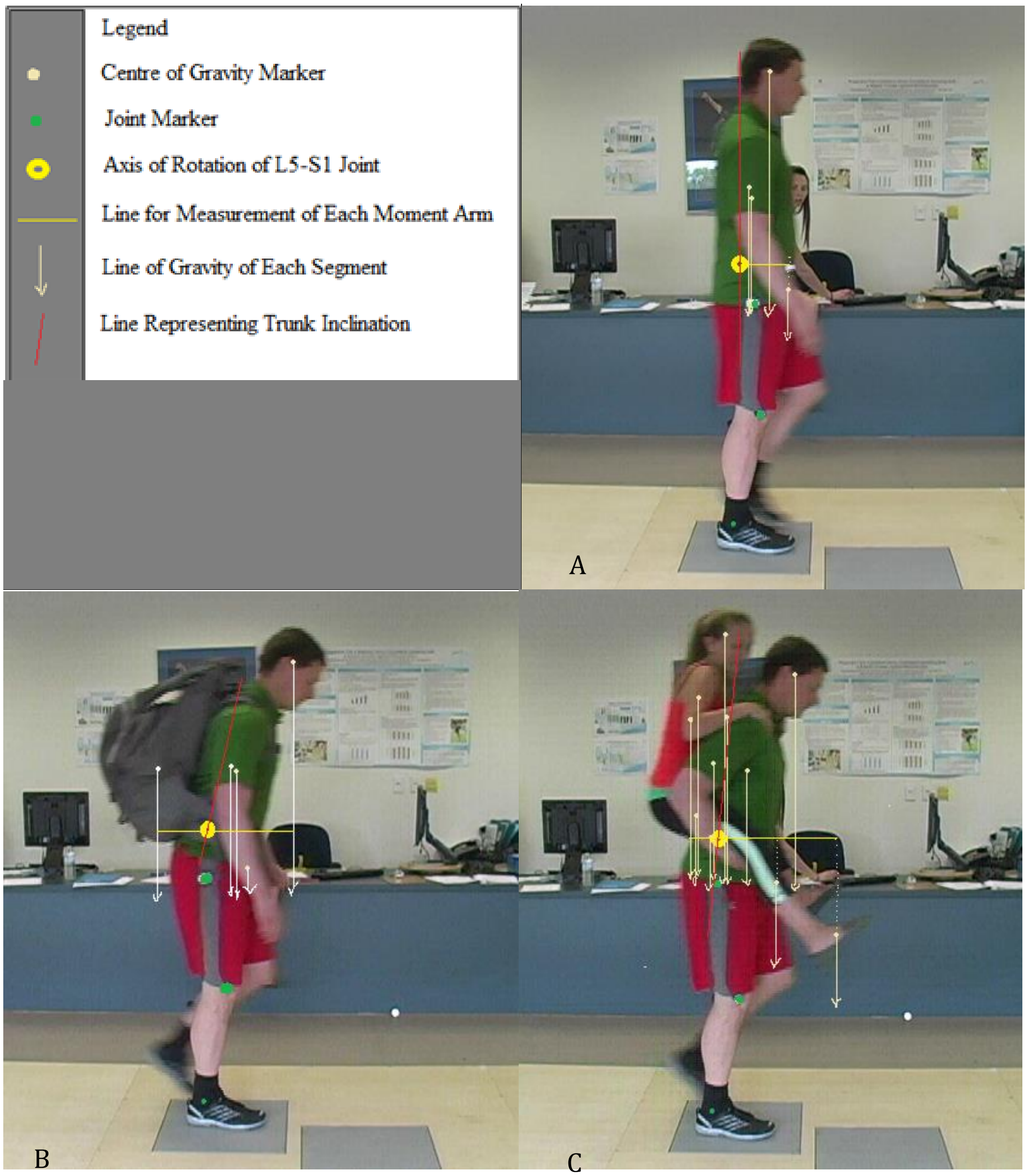


Figure 3.7: Measurements necessary to calculate lumbosacral load in each frame, over three conditions. The line of force of the muscle has been omitted. A represents the unloaded condition, B represents the backpack condition, C represents the piggyback condition.

The body was visually divided into segments connected at joint centres. Only the segments whose weight vectors acted superiorly to the lumbosacral joint centre, creating a moment at the lumbosacral joint, were included. The present model considered four body segments on the carrier and seven body segments on the passenger, each segment represented by its centre of mass location. The head, trunk, upper arm, and lower arm and hand segments were used for both the carrier and passenger. The thigh, lower leg, and foot segments were also used for the passenger calculations. These segments are illustrated in Figure 3.6. The force of gravity acting on these segment masses is illustrated in Figure 3.7 and is illustrated for backpacking more specifically in Figure 3.8.

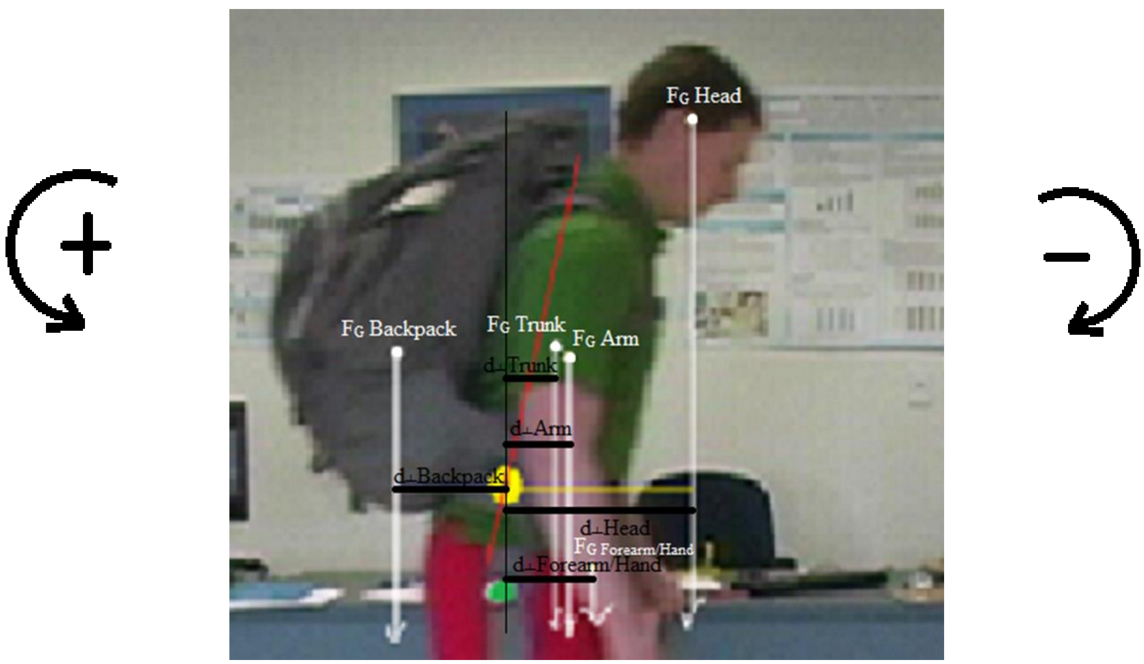


Figure 3.8: Segment weights and moment arms for a backpack example. The centre of gravity of each segment is represented by a white circle. The line of the force of gravity on each body segment is in white, and each corresponding moment arm is in black. The lumbosacral axis of rotation is located at the yellow dot.

Free body diagrams, like the one seen in Figure 3.9, illustrate various forces about the lumbosacral joint and were the basis for the static lifting model (Khoo et al., 1995). These forces were: the resultant muscle force, the force of gravity acting on each segment mass, the lumbosacral compressive joint reaction force, and the lumbosacral shear joint reaction force.

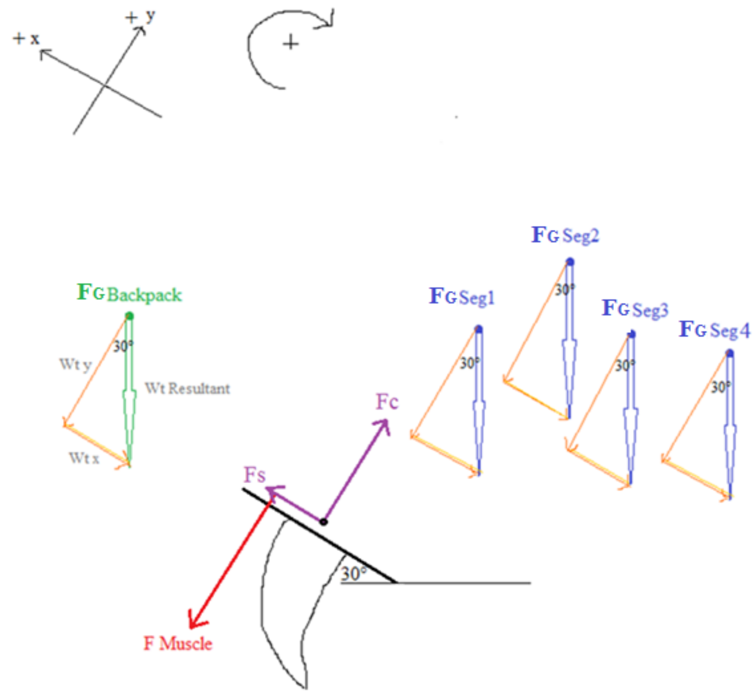


Figure 3.9: A Free Body Diagram: representing the forces present in the backpack condition. F_G represents the force of gravity (weight), or the product of the mass segment and the acceleration of gravity. Seg*n* represents each *n* segment. The black circle represents the axis of rotation of the lumbosacral joint. The orange lines represent the x and y components of the weights of each segment mass.

The direction and location of the summed pull of the lumbar spine erector muscles has been estimated and included as one vector representing all lumbar spine extensor muscles, as seen in Figure 3.9. Although most muscle fascicles do not exert a purely sagittal force (Bogduk et al., 1992), the calculated muscle torque and force will be a reflection of the amount of force that each muscle exerts in only the sagittal plane. The line of muscle force was assumed to be perpendicular to the surface of the lumbosacral joint. The resultant muscle moment and force were expected to represent the erector spinae muscle moment and force as opposed to those being exerted by the abdominal muscles, but this was determined by the positive or negative sign of the resultant torque. With backward rotation representing the positive direction, erector spinae muscle force was considered positive and abdominal muscle force was considered negative.

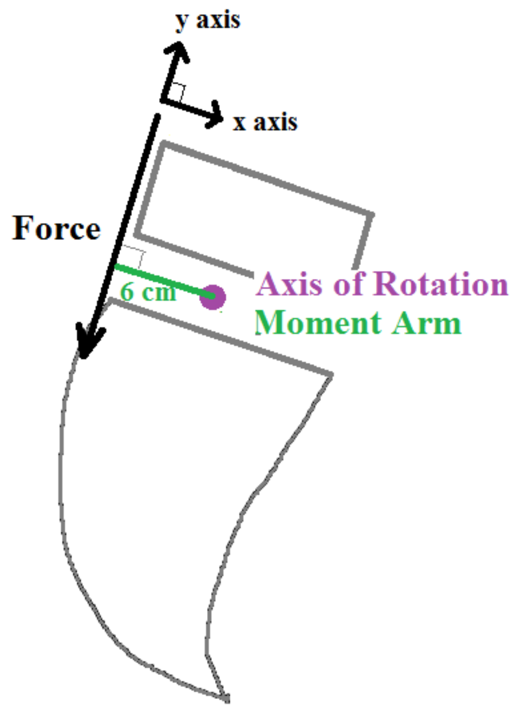


Figure 3.10: The line of pull and moment arm of the resultant of the summed extensor muscles at the lumbosacral Joint.

The moment arm of the resultant muscle force was parallel to the surface of the sacrum as it was perpendicular to the line of the muscle force. Based on previous estimations of the length of the resultant extensor muscle moment arm (Cholewicki et al., 1991; Hall, 2007), six centimeters was used for the resultant muscle moment arm length. This moment arm is represented by the green line in Figure 3.10.

The torque caused by each body segment and/or the backpack was estimated in order to balance the overall torque and solve for the resultant muscle torque. Because torque is the product of a force and its moment arm, the length of the moment arm for each segment weight was estimated. This was done by measuring the horizontal distance from each line of gravity to the axis of rotation of the lumbosacral joint, as represented by the black lines in Figure 3.8. The weight of each segment was found by multiplying the total body weight in Newtons by the respective body segment percentage in Table 3.1 on page 96.

The lines of the force of gravity acting on each applicable body segment were extended downward from each body segment's centre of gravity. The location of the centre of gravity of each body segment was determined on-screen in Dartfish, using the same method described for applying the beige tape markers at the start of this chapter (Hall, 2012). For the piggyback trials, this applied also to the passenger body segments. For the backpack trials, the line of the force of gravity extended downward from the centre of gravity of the backpack, which was located according to the method described at the start of this chapter.   The force of gravity acting on each segment was broken into y and x components to allow the summation of all forces in each the y axis and x axis. Forces perpendicular to the surface of the sacrum were considered parallel to the y axis and forces parallel to the surface of the sacrum were considered parallel to the x axis. The surface of the sacrum was assumed to be 30 degrees to the horizontal (Levangie & Norkin, 2005; Nordin & Frankel, 2012).

The compressive and shear joint reaction force vectors were drawn perpendicular and parallel to the joint surface respectively. They were calculated as the sums of the y and x axis forces respectively when each was set equal to zero. Because the compressive and shear components of the joint reaction force act at the centre of rotation of the joint, they had moment arms of zero, produced zero torque at the lumbosacral joint, and were not necessary in the first of the three static model equations, which balanced the moments at the lumbosacral joint (Chaffin et al., 1999).   With the constant sacral angle used in this model, the x components of the backpack and body segment weights did not alter and the shear force calculation stayed essentially the same for the same participant over any trials where they carried an equal weight of load. The compressive joint reaction force varied with trunk flexion angle because the muscle force, which was parallel to it, varied.

Calculating Lumbosacral Loads – Muscle Torque and Muscle Force

Three equations, all based on the assumption of static equilibrium, were used to calculate the lumbar load variables. The first equation balanced the moments about the lumbosacral joint. This provided the muscle torque value, which was divided by the muscle moment arm of six centimeters to equal the muscle force value. The second equation added the forces in the y direction to provide the compressive joint reaction force and the third equation added the forces in the x direction to provide the shear joint reaction force.

Because the piggyback condition had many mass segments and therefore many more terms in the equation, and the unloaded condition was lacking a force producing an extensor

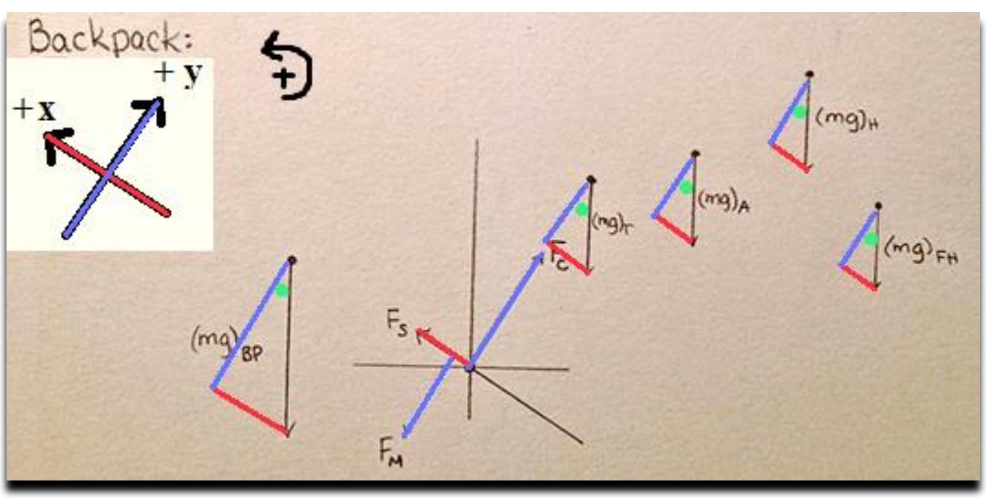


Figure 3.11: Free body diagram for the backpack condition. mg represents the force of gravity and the subscript represents the segment: T, A, H, and FH represent the trunk, arm, head, and forehead. The blue lines highlight the forces in the y dimension, the red lines highlight the forces in the x dimension, and the green dots represent an angle of 30 degrees. All other symbols are as used previously.

torque, the backpack condition was used to clearly illustrate these equations. A free body diagram representing the backpack condition is illustrated in Figure 3.11. The other two conditions are represented in Figure 3.12. The equation for torque was seen previously in Equation 1 in Chapter I, of which Equation 5 is more specific.

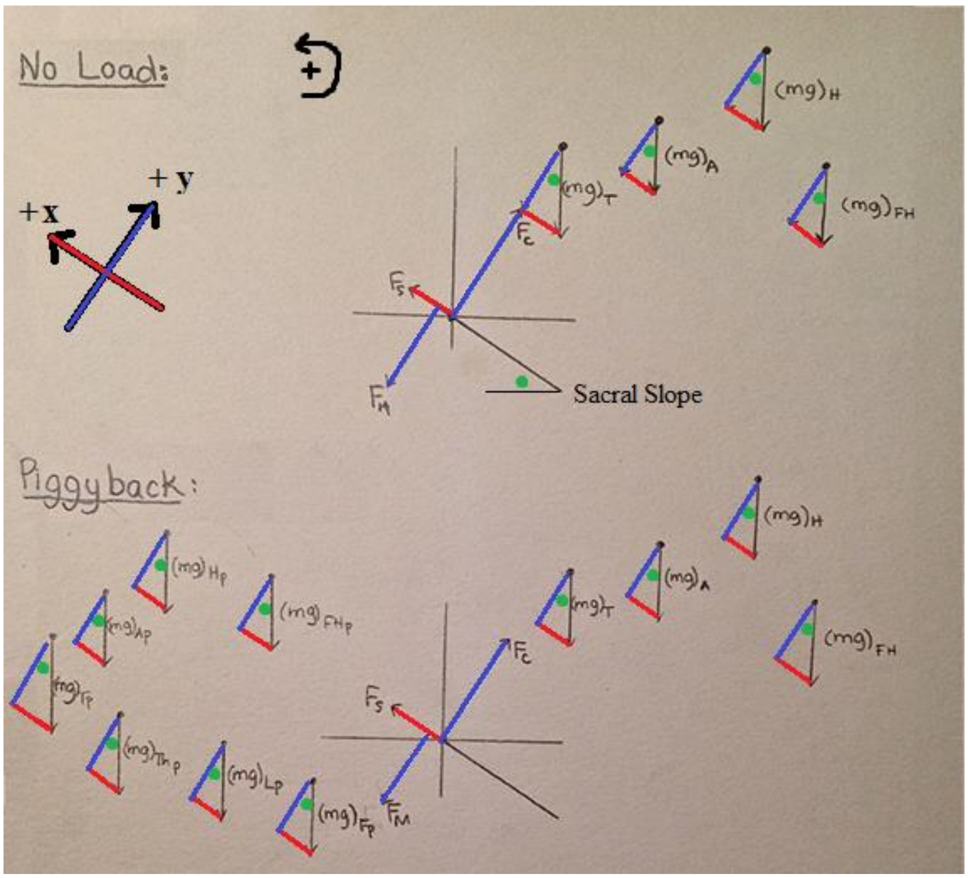


Figure 3.12: Free body diagrams for the unloaded and piggyback conditions. This is an elaboration of the previous figure to provide an idea of how this model can be applied to the unloaded and piggyback conditions. However, only the backpack condition will be calculated below. Subscript p represents passenger segments; Th, L, and F represent thigh, leg, and foot segments.

Equation 5: $T(\text{segment}) = [F_G(\text{segment}) \times d_{\perp}(\text{segment})]$

where T represents torque, F represents force, and d⊥ represents the moment arm of the force. Used to determine the magnitude of torque produced by any body segment

Resultant Muscle Torque Equation

Using a static model assumed equilibrium; the net forces and torques in two sagittal plane directions were summed to zero, and unknown values were determined. First, the sum of the moments at the lumbosacral joint was set to zero, summing all of the torques present as seen in Equation A (Benedek & Villars, 1973; Chaffin et al., 1999; Hall, 2007, Nordin & Frankel, 2012). Therefore, zero was equal to the sum of: the erector spinae muscle force times its moment arm, the weight of each body segment times its respective moment arm, and the weight of the backpack load times its moment arm (Chaffin et al., 1999):

Equation A:

$$\Sigma T = 0$$

$$\Sigma T = 0 = T_m + \Sigma T_{F\ posterior\ to\ axis} - \Sigma T_{F\ anterior\ to\ axis}$$

$$\Sigma T = 0 = T_m + (F_{G\ BP})(d_{\perp BP}) - (F_{G\ H})(d_{\perp H}) - (F_{G\ T})(d_{\perp T}) - (F_{G\ A})(d_{\perp A}) - (F_{G\ FH})(d_{\perp FH})$$

where F_G represents the force of gravity (mg or mass times the acceleration of gravity), and as a subscript: m represents muscle, H represents the head segment, T represents the trunk segment, A represents the upper arm segment, FH represents the forearm and hand segment, and BP represents the backpack. Solve for the unknown: extensor torque (Tm)
(Benedek & Villars, 1974, Chaffin et al., 1999; Hall, 2007; Nordin & Frankel, 2012)

Once this muscle torque (T_m) was calculated, the muscle force (F_m) was determined by dividing the torque by the estimated six centimeter length of the moment arm ($d_{\perp m}$), as seen in Equation 1.

Equation 1:

$$T = F\ d_{\perp}$$

$$Tm = (F_m)(d_{\perp m})$$

$$Tm / (d_{\perp m}) = F_m$$

Solve for the unknown: resultant muscle force (F_m)
(Benedek & Villars, 1974, Chaffin et al., 1999; Hall, 2007; Nordin & Frankel, 2012)

Muscle torque and muscle force sample calculations have been provided below.

For Condition B, in which the load was carried in the form of a backpack (BP):

$$\sum T = 0 = T_{muscle} + T_{BP} - T_{head} - T_{trunk} - T_{arms} - T_{\substack{forearm \\ hand}}$$

$$\sum T = 0 = T_{muscle} + (284.49)(.19) - (60.92)(.16) - (424.25)(.06) - (44.02)(.14) - (37.99)(.25)$$

$$\sum T = 0 = T_{muscle} + (54.05) - (9.75) - (25.46) - (6.16) - (9.50)$$

$$\sum T = 0 = T_{muscle} + 3.18$$

Therefore: T_{muscle} = – 3.18 Nm, *and:*

$$F_{muscle} = T_{muscle} / d_{\perp}$$

$$F_{muscle} = -3.18 / 0.06$$

$$F_{muscle} = -53.00\ N$$

Summary: Because the value is negative, the abdominal muscles were estimated to be providing 53 N of force to keep the trunk from rotating in a backward direction at maximum trunk flexion while walking with a 284.49 N backpack.

Compressive and Shear Joint Reaction Force Equations

The same assumption of equilibrium was held while calculating the compressive and shear lumbosacral joint reaction force components. Therefore, the sum of the forces in the y dimension was set equal to zero and the sum of the forces in the x dimension was set equal to zero and both unknown values, the compressive component and shear component respectively, were determined. This included the components of the weight of the backpack and each segment of the passenger if they were involved. Figure 3.12 illustrates the angles of the force vectors with respect to the y and x axes, which are perpendicular and parallel to the surface of the sacrum. This figure aids in understanding the trigonometric equations needed to represent each component of the force of gravity on each body segment. The force of gravity acting on each body segment in the y axis is equal to the total weight of that body segment multiplied by the cosine of 30 degrees and the force of gravity acting on each body segment in the x axis is equal

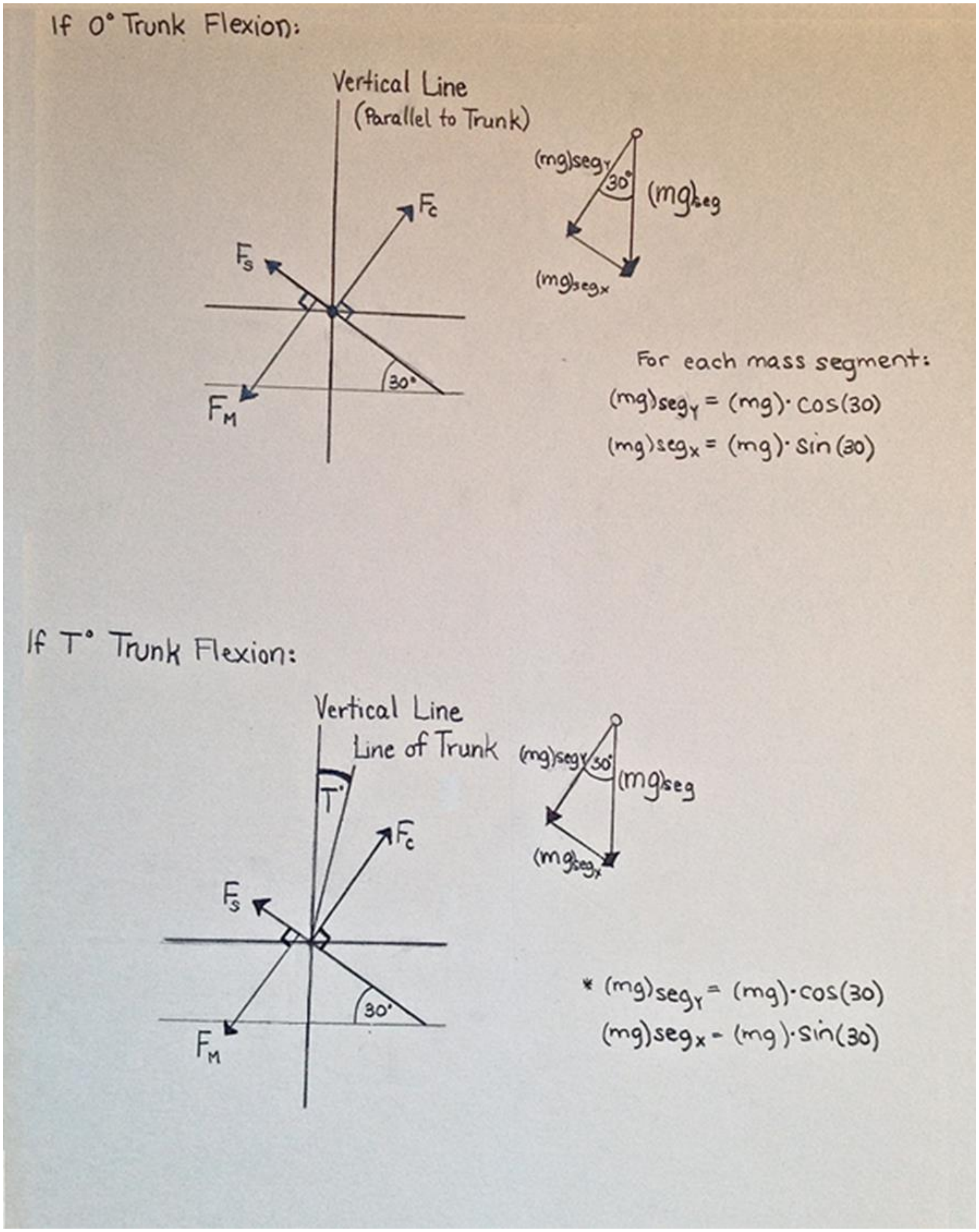


Figure 3.13: Trigonometry was used to determine the value of the component of the force of gravity acting on each body segment in both the x and y axes.

to the total weight of that body segment multiplied by the sine of 30 degrees. Figure 3.13 illustrates that the compressive and shear joint reaction force values do not change in this model during trunk flexion.

For the y axis, the sum of the forces perpendicular to the superior sacral surface and parallel to the compressive joint reaction force was equal to zero. These forces were: the compressive joint reaction force, the resultant muscle force, and the y axis components of the weight of each body segment and the backpack. This calculation is seen in Equation B.

Equation B: $\sum F_y = 0$

$$\sum F_y = 0 = F_C - F_m - F_{BP_y} - \Sigma(F_{weight\ of\ each\ segment_y})$$

$$0 = Fc - Fm - \cos\theta\,(\mathrm{FG})BP - \cos\theta\,(\mathrm{FG})H - \cos\theta\,(\mathrm{FG})T - \cos\theta\,(\mathrm{FG})A - \cos\theta\,(\mathrm{FG})FH$$

where $\sum F_y$ represents the sum of the forces in the y axis, F_C represents the compressive lumbosacral joint reaction force, F_m represents the muscle force, F_{BP_y} represents the force of the weight of the backpack, if it is worn, in the y axis, $\Sigma(F_{weight\ of\ each\ segment_y})$ represents the addition of the weight of each body segment of the participant and, if present, the passenger, in the y axis, θ represents the angle between the superior surface of the sacrum and the horizontal, or 30° (Chaffin et al., 1999). *Solve for the only unknown (Fc).*
(Chaffin et al., 1999; Levangie & Norkin, 2005; McGill & Norman, 1985; Nordin & Frankel, 2012)

The above equation was solved for the only unknown, the compressive joint reaction force (Fc). Then, the same calculations were performed for the x axis, as seen in Equation C:

Equation C: $\sum F_x = 0$

$$\sum F_x = 0 = F_s - F_{BP_y} - \Sigma(F_{weight\ of\ each\ segment_x})$$

$$0 = Fs - \sin\theta(FG)BP - \sin\theta(FG)H - \sin\theta(FG)T - \sin\theta(FG)A - \sin\theta(FG)FH$$

where $\sum F_y$ *represents the sum of the forces in the x axis,* F_s *represents the shear lumbosacral joint reaction force, ,* F_{BP_x} *represents the force of the weight of the backpack, if it is worn, in the x axis,* $\Sigma(F_{weight\ of\ each\ segment_x})$ *represents the addition of the weight of each body segment of the participant and, if present, the passenger, in the x axis. Solve for the only unknown (Fs).*
(Chaffin et al., 1999; Levangie & Norkin, 2005; McGill & Norman, 1985; Nordin & Frankel, 2012)

The above equation was solved for the only unknown, the shear joint reaction force (Fs). The resultant joint reaction force (JRF) was determined by using the shear and compressive components in Pythagoras Theorem:

Equation 6: $\mathrm{JRF} = \sqrt{(Fc^2+Fs^2)}$ (Khoo et al., 1993)

where JRF represents the resultant lumbosacral joint reaction force.

Compressive, shear, and resultant lumbosacral joint reaction force calculations have been provided below. The calculation has been continued for the participant during the backpack condition (2012), in which the load was carried in the form of a backpack (BP).

$$\sum F_y = 0$$
$$\sum F_y = 0 = F_C - F_m - F_{BP_y} - \Sigma(F_{weight\ of\ each\ segment_y})$$
$$0 = Fc - Fm - \cos\theta\,(\mathrm{FG})BP - \cos\theta\,(\mathrm{FG})H - \cos\theta\,(\mathrm{FG})T - \cos\theta\,(\mathrm{FG})A - \cos\theta\,(\mathrm{FG})FH$$
$$0 = Fc - 53.00 - 246.89 - 52.87 - 368.18 - 38.20 - 32.97$$
$$Fc = 792.11\ \mathrm{N}$$

$$\sum F_x = 0$$
$$\sum F_x = 0 = F_s - F_{BP_y} - \Sigma(F_{weight\ of\ each\ segment_x})$$
$$0 = Fs - \sin\theta(FG)BP - \sin\theta(FG)H - \sin\theta(FG)T - \sin\theta(FG)A - \sin\theta(FG)FH$$
$$0 = Fs - 141.36 - 30.27 - 210.80 - 21.87 - 18.88$$
$$Fs = 423.18\ \mathrm{N}$$

$$\mathrm{JRF} = \sqrt{(Fc^2+Fs^2)}$$
$$\mathrm{JRF} = \sqrt{(792.11^2 + 423.18^2)}$$
$$\mathrm{JRF} = 898.06\ \mathrm{N}$$

Summary: the compressive joint reaction force was at least 792.11 N, the shear joint reaction force was at least 423.18 N, and the resultant joint reaction force was 898.06 N at support flexion while walking with a 284.49 N backpack.

Summary of the Steps for Lumbar Load Data Calculations

An Excel spreadsheet was created to allow measured values to be entered for each trial in order to calculate each of the lumbar load variables. Then, the maximum trunk flexion Dartfish key position was located for each trial. The lumbosacral joint was marked. Each applicable body segment length was measured and multiplied by the appropriate proportion found in Table 3.1 on page 96; the resulting distance was measured distally from its proximal joint and marked as the segment centre of gravity. The backpack or passenger segment centres of gravity were marked. Vertical lines were drawn through each centre of gravity. The horizontal distance from each centre of gravity to the lumbosacral joint was measured and recorded in Excel. Excel was programmed to produce values for the lumbar load variables.

Other Torque Calculations

Additional variables were added to better compare the lumbosacral joint torques between conditions. Resultant torque by load only, total magnitude of torque by load only, negative passenger torque, positive passenger torque, percent negative passenger torque, percent positive passenger torque, torque produced by body of carrier, total negative torque, total positive torque, total resultant torque, total magnitude of torque, percent negative total torque and the percent positive total torque were calculated. Several measurements that were already used in the Excel formula that calculated the total resultant torque, the first step of the lumbar spine model, were added to calculate these new variables.

The lumbosacral joint torque produced by the external load, during piggybacking and backpacking, was summed and documented as the *resultant torque by load only*. For piggybacking, all of the passenger body segment torques were added; these were the torques caused by the head, trunk, upper arms, forearms and hands, thighs, lower legs, and feet and some of these values were negative. For backpacking, the torque produced by the backpack was already calculated and that value was used for resultant load torque; every backpack torque value was positive.

Whereas the resultant load torque variable summed the positive and negative torque values, the *total magnitude of torque by load only* variable summed the absolute values of the torques caused by each body segment, as listed above. Because the backpack only caused positive torque, the values for the backpacking condition are the same for this and the previous variable. However, the total magnitude of the torques caused by the passenger's body segments was greater than the resultant load torques as seen in the previous variable.

The positive and negative components of the total magnitude of torque caused by the passenger segments were also compared and a percentage was calculated for each. These values were expressed as *negative passenger torque*, *positive passenger torque*, the *percent negative passenger torque* and the *percent positive passenger torque.*

Separate from the torque caused by any external load, the resultant torque produced by the body segments of the carrier was calculated for the *torque produced by carrier* variable. This included the lumbosacral torques caused by the head, trunk, upper arms, and forearms and hands of the carrier. All of these values were negative.

The *total negative torque* variable was the sum of all of the negative torques caused by the body segments of the carrier and/or passenger, which were the torques caused by any segment anterior to the lumbosacral joint. The backpack always created positive torque and was therefore never included in this variable. For the no load and backpack conditions, this value was equal to the torques caused by the body segments of the carrier. For the piggyback condition, this was the sum of the torques produced by the body segments of the carrier plus the torque produced by any passenger body segment with a negative value.

The *total positive torque* variable was the sum of all of the positive torques caused by the body segments of the passenger or backpack; the centre of every body segment of the carrier was anterior to the lumbosacral joint, always producing negative torque. For the no load condition, this value would represent the positive torque caused by the body of the carrier but no carrier body segment was found to produce overall positive torque; all values for the no load condition were equal to zero. For the piggyback condition, the total positive torque value represented the total amount of torque caused by parts of the body of the passenger that were posterior to the lumbosacral joint of the carrier. For the backpack condition, this value represented the torque caused by the backpack.

The *total resultant torque* was the summation of the total negative torque and the total positive torque variables. Because the positive torque was estimated at zero during the no load condition, the total torque during the no load condition was equal to its total negative torque. This summation produced new values for both of the loaded conditions. The *total magnitude of torque* was calculated by summing the absolute value of the total negative torque to the total positive torque. Both the total negative torque and total positive torque variables were also calculated as a percentage of the total magnitude of torque, known as the *percent of total torque produced by negative torque* and the *percent of total torque produced by positive torque*.

Kinetic Variables – Vertical Ground Reaction Forces

The vertical ground reaction forces were graphed over time throughout the duration of right foot stance, and specific peaks were identified. These graphs allowed for the vertical ground reaction force patterns to be visually evaluated. Three specific events of the ground reaction force curve were recorded and compared between each other and between each condition. These were the highest magnitude of the first impact peak, the lowest magnitude of the dip between peaks, and the highest magnitude of the second peak of the vertical ground reaction force. They were found by examining the force plate Excel spreadsheet; the first greatest value before the values decreased, the lowest value before the values increased again, and the last greatest value before the values decreased to zero were the values that marked these events.

The ground reaction force values were used both directly and with normalization, which was calculated by dividing the ground reaction force in Newtons by the weight of the entire system in Newtons. The normalized unit was *TW or "times total weight". These three variables were also addressed with respect to each increase from the no load value to the loaded value. Note that both the piggyback load and the backpack load were 284.49N. For each variable, the no load vGRF was subtracted from the piggyback vGRF. This value was divided by 284.49 and multiplied by 100 to determine the percentage of the added load that manifested as added vGRF. This calculation was repeated for the backpack condition.

A Butterworth filter was tested on the pilot study data, based on recommendations from an article on data filtering (Bogert & Koning, 1996). It did not substantially affect the data; for example, it converted one maximum value from 1268.147 N to 1268.253 N (Graham, 2012).

Because of the results of the 1996 paper and simple experimentation by the present researcher, the decision was made not to use a filter on the force plate data in this project.

Gait Variables

Several gait parameters were investigated, including walking velocity, walking cadence, average step length, right to right stride length, left to left stride length, right foot stance time, single stance time, and double stance time. The 0.47m wide force plate was used as the distance calibration within the plane of the carrier during Dartfish analysis. Because it was more important to record a natural gait pattern than force at a constant speed, the participants were asked to walk at a natural pace and their walking velocity was measured onscreen. The distance across the screen was measured and divided by the time it took the centre of the head to enter and exit the screen; this was recorded as the walking velocity in meters per second.

Walking cadence was also measured and recorded, but in steps per minute. The time it took from left heel strike to left heel strike was measured and inserted into the formula below:

Conversion Formula: (60s) * (2 steps) ÷ (# seconds it takes for 2 steps) = Cadence in steps/min

Right to right stride length, and left to left stride length were measured using the distance measuring tool in Dartfish. For right to right stride length, the location of right heel strike was marked on screen before the video was moved forward and a second right heel strike was seen and marked. The distance between the two markers was converted using the force plate conversion factor as previously mentioned. The same procedure was repeated for left to left stride length. Once these two distances were recorded, an Excel formula was developed that added them and divided the total by four to determine another variable: the average step length.

Right foot stance time was measured from the Excel spreadsheet data collected from the force plate at 200 frames per second. The number of frames in which the right foot was in contact with the ground was counted and multiplied by 0.005 of a second to provide the total amount of time that the right foot was in stance.

Single stance time and double stance time were measured onscreen using the times provided by Dartfish. First, the number of frames during one gait cycle where only one foot was in contact with the ground and the number of frames where both feet were in contact with the ground were counted by scrolling frame by frame. Because the video was filmed at 30 frames per second, those values were then multiplied by 0.033 to determine the total duration of single stance phase and the total duration of double stance phase during one gait cycle in seconds.

STATISTICAL ANALYSIS

Each variable was compared amongst the three conditions using a Repeated Measures Analysis of Variance (ANOVA) test utilizing SPSS version 21 software. Mean values were determined for each participant for each variable, and these values were compared amongst the three conditions of no load, piggybacking, and backpacking. An alpha value of 0.05 was used. When results were significant, a Bonferroni post-hoc test was used to determine which conditions were statistically different from each other. As previously mentioned in Chapter I, the null and alternative hypotheses are stated below.

Null Hypothesis: $\mu_{NL} = \mu_{PB} = \mu_{BP}$ where μ is the mean of the population of conditions UL, PB and BP.

Alternative Hypothesis: μ_{NL}, μ_{PB}, and μ_{BP} are not all equal. (D'Agostinoet al., 2006)

When a comparison was not significant, the null hypothesis was accepted and there was no difference in that variable between the no load, piggyback, and/or backpack conditions. When a statistically significant result occurred and the null hypothesis was rejected, a Bonferroni post hoc test was performed to determine which conditions were different from each other.

CHAPTER IV

RESULTS

KINEMATIC VARIABLES

Results for all kinematic variables are listed in Table 4.1. Every variable exhibited significant differences between at least two conditions.

Table 4.1: Kinematic variable means, standard deviations, and pairwise p-values across conditions.

Variable\Condition	No Load	Piggyback	Backpack	Bonferroni p-value		
				NL/BP	NL/PB	BP/PB
Max Trunk Flex Angle (°)	0.9±1.1	9.7±3.1	12.4±2.3	**.000**	**.000**	**.002**
Min Trunk Flex Angle (°)	-0.1±1.3	7.3±3.2	8.2±3.2	**.000**	**.000**	.927
Trunk Flex-Ext ROM (°)	2.2±1.1	2.3±1.1	4.2±1.7	**.002**	1.000	**.003**
Max Hip Flex Angle (°)	27.5±2.9	35.5±3.3	38.4±3.5	**.000**	**.000**	.050
Max Knee Flex Angle (°)	17.8±8.9	22.6±3.0	24.2±3.7	**.026**	.126	**.026**

- A Bonferroni post hoc test was used if repeated measures ANOVA determined significance
- Statistically significant p-values are bolded, α = 0.05

Maximal Trunk Flexion Angle

The maximum angle of trunk flexion while walking was different between all three conditions, based on a repeated measures ANOVA with a Bonferroni post hoc test.

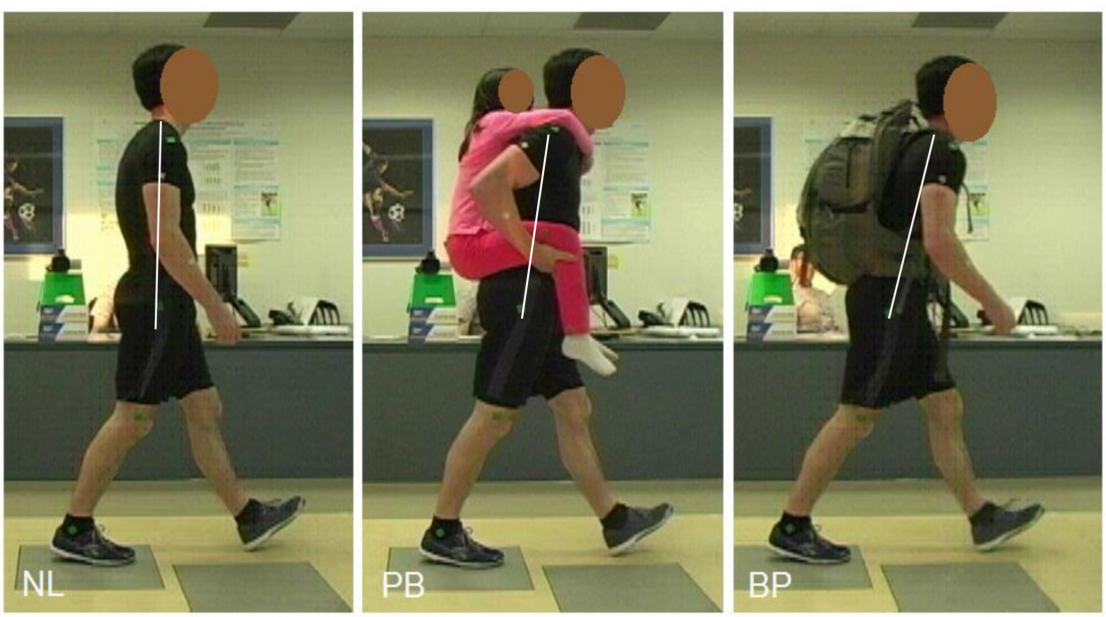


Figure 4.1: Still frames at maximum trunk flexion by load condition.

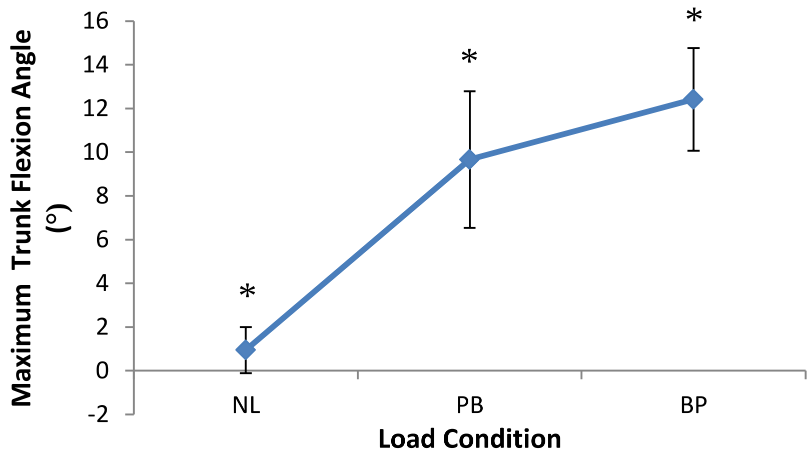


Figure 4.2: Maximum trunk flexion angle by load condition. *At α = 0.05, the maximum angle of trunk flexion was significantly different between each pair of load conditions.

Trunk flexion increased from no load to piggybacking (p=0.000), from no load to backpacking (p=0.000), and from piggybacking to backpacking (p=0.002) as demonstrated visually in Figure 4.1 and graphically in Figure 4.2. The trunk was most vertical when carrying no load (0.9±1.1°), flexed when piggybacking a person (9.7±3.1°), and the most flexed when carrying an equally-weighted backpack (12.4±2.3°).

Other Kinematic Variables

The angle of minimum trunk flexion increased from carrying no load to carrying each load, but did not differ between loaded conditions. A greater range of trunk flexion-extension motion was exhibited while backpacking than either of the other two conditions, which did not differ from each other. Trunk angles and range of motion can be considered while referring to Figure 4.3 and Figure 4.4.

The angle of relative maximum hip flexion increased from no load to each of the loaded conditions. Because hip flexion angle was measured between the line extended from the trunk and the line of the thigh, this angle reflected the change in trunk flexion. When the influence of trunk flexion on hip flexion angle was investigated by subtracting maximal trunk flexion angle

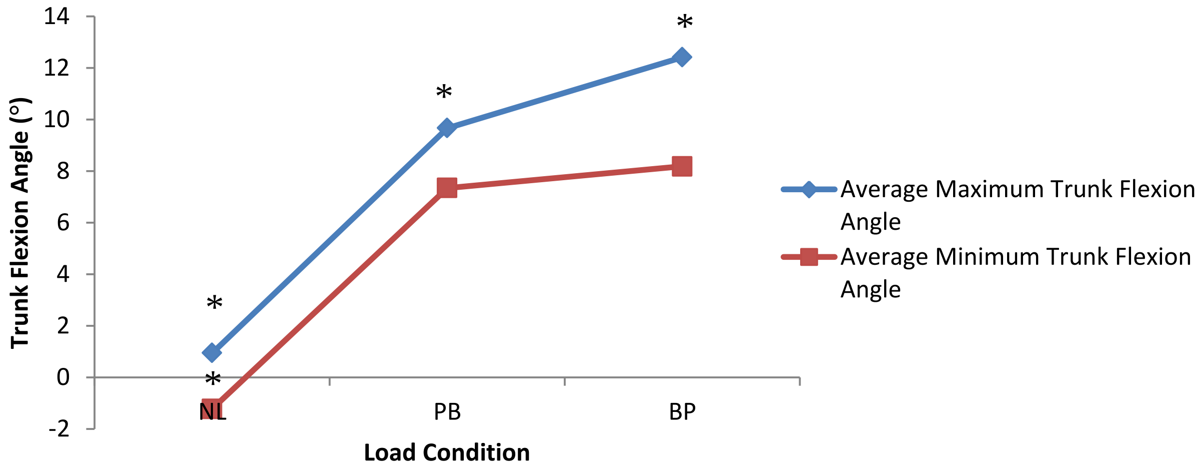
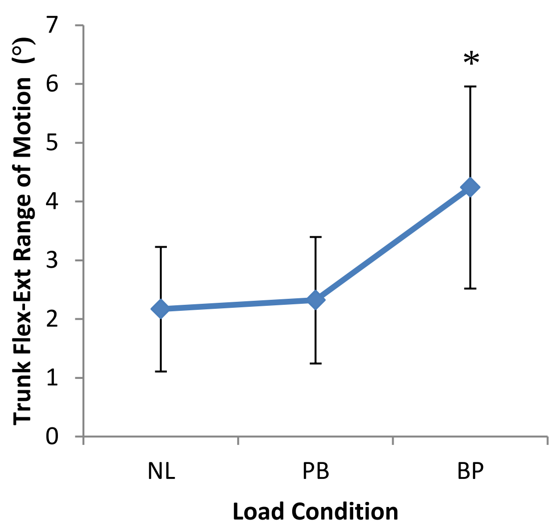


Figure 4.3: A comparison of the average maximum and average minimum angles of trunk flexion while walking in each load condition. The difference between the two averages roughly represents the trunk flexion-extension range of motion. *At $\alpha = 0.05$, the maximum trunk flexion angle was different between all pairs of conditions while minimum trunk flexion angle was only different from no load to each loaded condition.



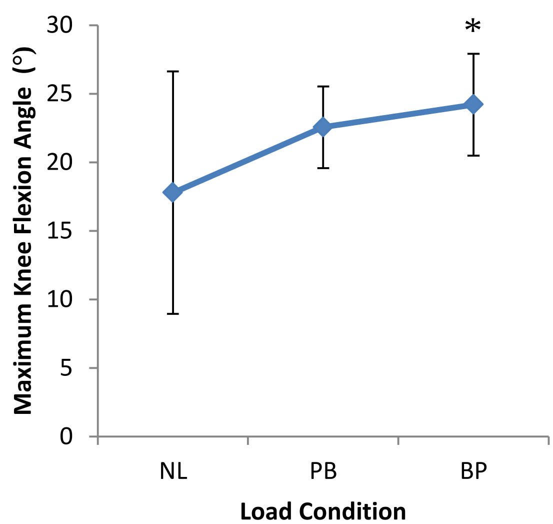


Figure 4.4: The trunk flexion-extension range of motion by load condition. *At $\alpha = 0.05$, the range of trunk motion was greater for backpacking than for each of the other two conditions.

Figure 4.5: Average maximum knee flexion angle by load condition. * At $\alpha = 0.05$, knee flexion angle was greater for backpacking than for each of the other two conditions.

from maximal hip flexion angle, the average absolute hip flexion angles from the vertical axis were approximately 27±3˚, 26±3˚ and 26±3˚ for no load, piggybacking and backpacking respectively. These values were estimates; the instants of maximum trunk flexion and maximum hip flexion were not necessarily simultaneous, but the trunk angles were usually similar. None of the above estimated absolute hip angle values were significantly different. Therefore, load condition did not determine absolute hip flexion angle, and the previous differences in relative hip angle were likely only a result of trunk flexion.

The angle of maximum knee flexion was greatest when carrying a backpack, which varied from maximum knee flexion angle when carrying a person and when carrying no load. The mean maximum knee angle for piggybacking was also greater than the mean for no load, but this difference was not significant. Maximum knee flexion angle across conditions is illustrated in Figure 4.5.

EXTERNAL LUMBOSACRAL TORQUE VARIABLES – LOAD TORQUES

Table 4.2: Lumbosacral torque means, standard deviations, and pairwise p-values across conditions.

Variable\Condition	Torque (Nm)			Bonferroni p-value		
	No Load	Piggyback	Backpack	NL/BP	NL/PB	BP/PB
Resultant Torque (T) by Load Only		16.68±4.81	54.69±6.67			**.000**
Total Magnitude of T by Load Only		33.42±3.33	54.69±6.67			**.000**
T by Body of Carrier	-27.66±6.42	-38.70±7.27	-48.73±8.75	**.000**	**.000**	**.000**
Total Resultant T	-27.66±6.42	-22.01±10.73	5.96±13.15	**.045**	**.000**	**.000**
Total Magnitude of T	27.66±6.42	72.28±5.58	103.41±8.33	**.000**	**.000**	**.000**
Total Negative T	-27.66±6.42	-47.15±7.62	-48.73±8.75	**.000**	**.000**	.760
Total Negative T as %	100.00±0.00	65.17±6.53	46.92±6.57	**.000**	**.000**	**.000**
Total Positive T	0.00±0.00	25.14±3.88	54.69±6.67	**.000**	**.000**	**.000**
Total Positive T as %	0.00±0.00	34.83±6.53	53.08±6.57	**.000**	**.000**	**.000**

- A Bonferroni post hoc test was used when the repeated measures ANOVA determined significance
- Instead, a Paired T-Test was used when only two conditions were compared
- Statistically significant p-values are bolded, $\alpha = 0.05$
- Backward rotation was considered positive

Every external lumbosacral torque variable was different between all three conditions except for the total amount of negative torque, which was similar between the two loaded conditions. Details of the load torque variables estimated by the lumbar spine model are listed in

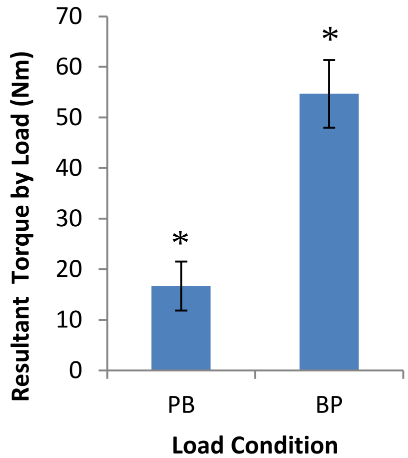


Figure 4.6: Resultant torque produced by load during piggybacking versus backpacking. * At α = 0.05, a greater resultant torque was produced by backpack than passenger.

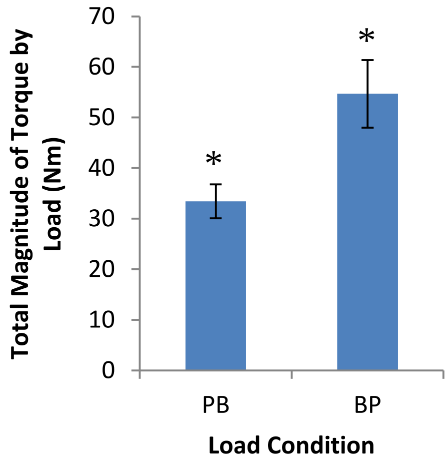


Figure 4.7: Total magnitude of torque produced by load during piggybacking versus backpacking. * At α = 0.05, a greater total magnitude of torque was produced by backpack than passenger.

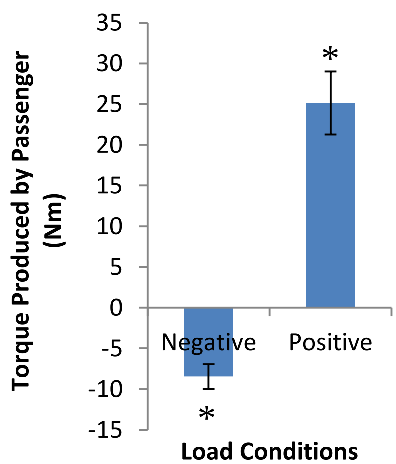


Figure 4.8: Torque produced by passenger load, divided into negative and positive components. * At α = 0.05, a paired t-test found the postive torques produced by the passenger load to be greater than the negative torques produced by the passenger load.

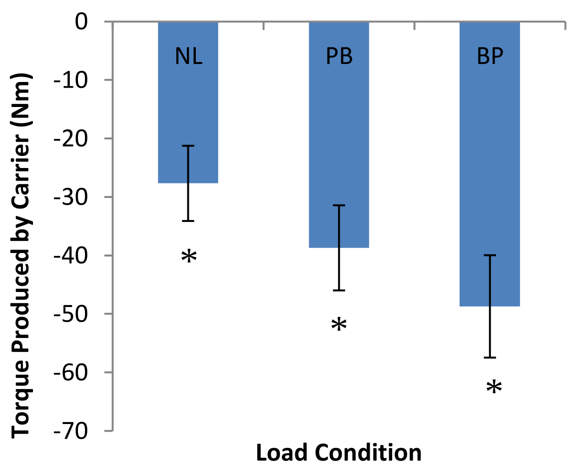


Figure 4.9: Torque produced by upper body segments of the carrier across load conditions. * At α = 0.05, the upper body segments of the carrier produced different forward torques between all conditions.

Table 4.2; backward rotation was considered positive. The torques caused solely by the load, as the passenger or the backpack, were investigated by Figures 4.6 and 4.7. Both the resultant torque and total magnitude of the torque caused by the load were statistically greater for the backpack than for the passenger. For the piggyback condition, the torques caused by the body segments of the passenger were divided into positive and negative torques. Approximately 25% (-8.45±1.51 N) of the total magnitude of torque caused by the passenger was negative, while approximately 75% (25.13±3.88N) of the passenger torque was positive. Overall, the load itself increased the tendency to rotate backward during piggybacking. This comparison is illustrated in Figure 4.8.

The torque produced by the upper body segments of the carrier statistically increased from no load to piggybacking to backpacking, as seen in Figure 4.9. The torque produced by the

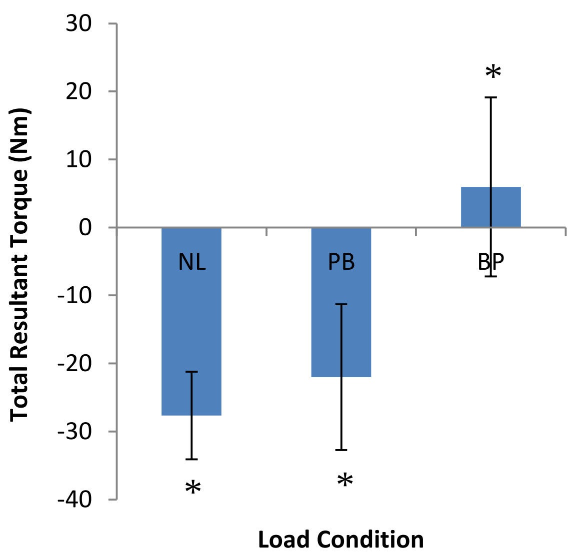


Figure 4.10: Total resultant torque demonstrated by each condition. * At α = 0.05, the resultant torque was significantly different between all conditions.

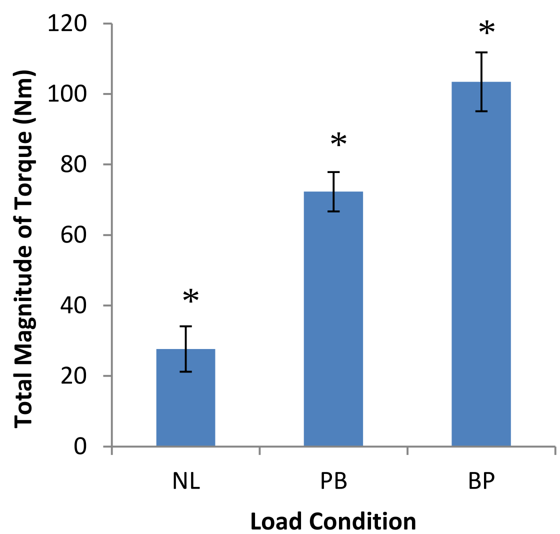


Figure 4.11: Total magnitude of torque demonstrated by each condition. * At α = 0.05, the total magnitude of torque was different between all conditions.

upper body segments of the carrier increased as the moment arms of each segment weight increased with trunk flexion angle, as seen from no load to piggybacking to backpacking.

The total resultant torque increased from no load to the piggyback condition to the backpack condition; both the no load and piggyback conditions demonstrated an overall tendency for the system to rotate forward while the backpack condition demonstrated a tendency for the system to rotate backward. This is illustrated in Figure 4.10. The total magnitude of the system torque also increased from no load to piggybacking to backpacking, seen in Figure 4.11.

The positive and negative components of the total resultant torque produced by the carrier and load, if one was carried, was divided into total negative and total positive torque values, as seen in Figure 4.12. The magnitude of negative torque increased from no load to piggybacking to backpacking. Figure 4.13 illustrates the total positive and total negative torque values as percentages of the total magnitude of system torque. Figures 4.12 and 4.13 illustrate that all torque produced when unloaded was negative; it was produced by the carrier's body.

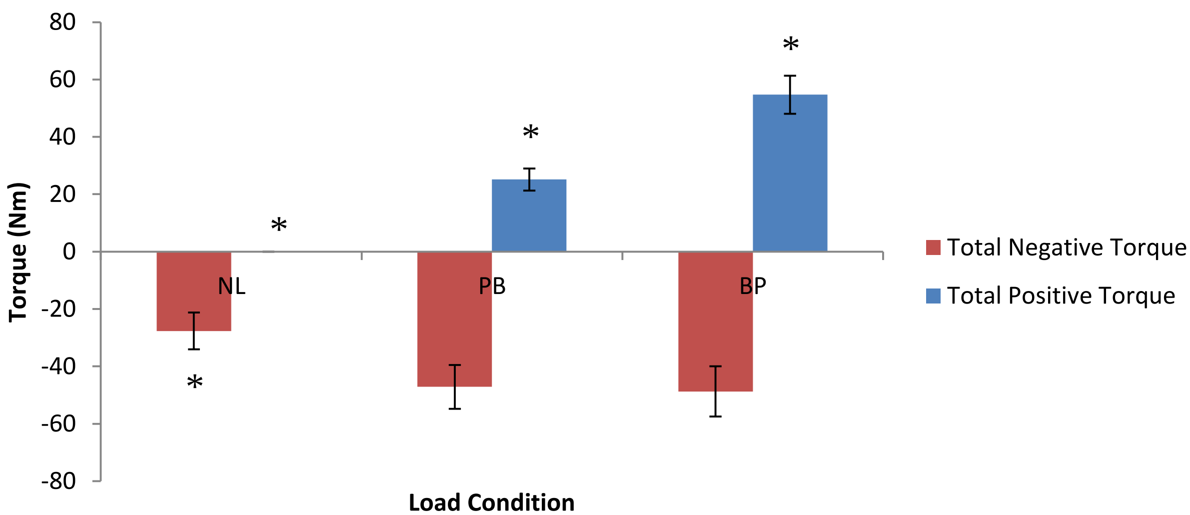


Figure 4.12: Total negative versus total positive torques demonstrated by each condition.
* At $\alpha = 0.05$, the total negative torque for no load was less than for the loaded conditions; the total positive torque was different between all conditions.

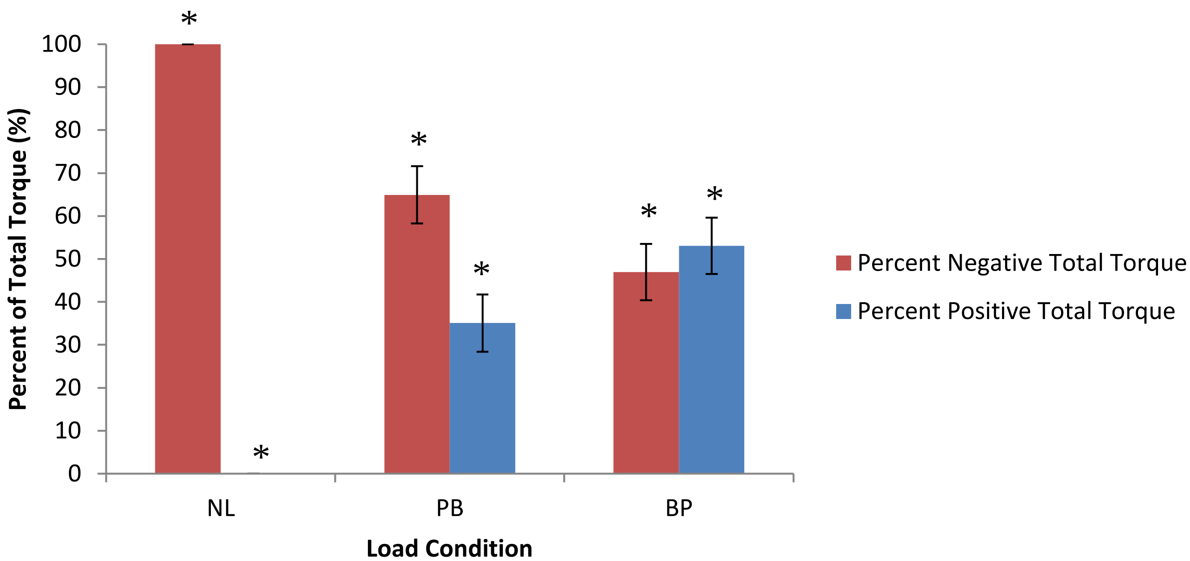


Figure 4.13: Negative and positive torques as a percentage of total magnitude of torque, as demonstrated by each condition. * At α = 0.05, the percentages were different between all conditions for both variables.

For piggybacking, the anterior body segments of both the carrier and passenger created more negative torque than the portion of the passenger that produced postive torque. Lastly, the positive torque produced by the backpack was close to but greater than the negative torque produced by the body segments of the carrier. When summed, the resultant torque described the overall tendency for rotation around the medio-lateral axis of the lumbosacral joint.

INTERNAL LUMBOSACRAL TORQUE AND FORCE VARIABLES

Table 4.3: Lumbar load means, standard deviations, and pairwise p-values across conditions.

Variable\Condition	No Load	Piggyback	Backpack	Bonferroni p-value NL/BP	NL/PB	BP/PB
Resultant Muscle Torque (Nm)	27.66±6.42	21.77±10.91	-5.96±13.15	**.000**	**.039**	**.000**
Resultant Muscle Force (N)	460.98±107.08	362.80±181.80	-99.40±219.39	**.000**	**.039**	**.000**
Compressive L5-S1 JRF (N)	14.49±79.57	355.40±150.27	821.62±181.02	**.000**	**.000**	**.000**
Shear L5-S1 JRF (N)	272.24±27.43	413.59±27.43	413.59±27.43	**.000**	**.000**	1.000
Resultant L5-S1 JRF (N)	287.08±30.50	581.84±82.97	923.50±155.22	**.000**	**.000**	**.000**

- A Bonferroni post hoc test was used when the repeated measures ANOVA determined significance
- Statistically significant p-values are bolded, α = 0.05

All estimated lumbar loads are listed in Table 4.3. As a reminder, the torque calculation relies on the balancing of torques, which only provides the resultant torque in one direction rather than the absolute torque in each direction. Because resultant torque magnitude must be less than absolute torque magnitude and the remaining lumbar load variables were calculated based on the resultant muscle torque, these variables were underestimations to some degree. Resultant muscle torque increased from the backpack condition to the piggyback condition to the no load condition. The resultant muscle torque was a negative value in the backpack condition and a positive value in the other two conditions; therefore, the trunk extensors produced greater torque than the trunk flexors during the piggybacking and no load conditions, but the trunk flexors dominated during backpacking. The resultant muscle torque was very similar between piggybacking and carrying no load, with a Pearson correlation coefficient of 0.8. Resultant muscle torque averages can be seen in Figure 4.14.

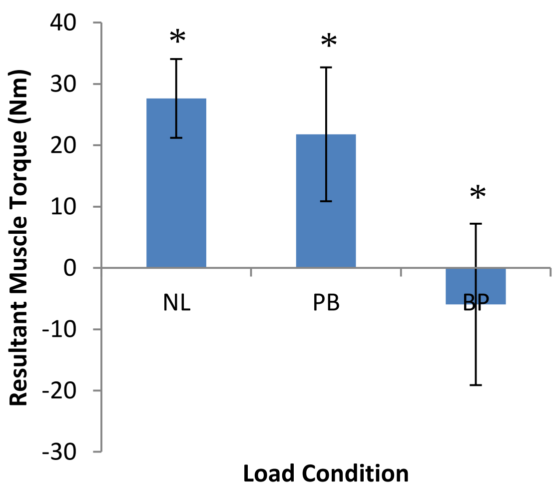


Figure 4.14: Estimated resultant muscle torque by load condition. *At α = 0.05, significant differences were seen between all conditions.

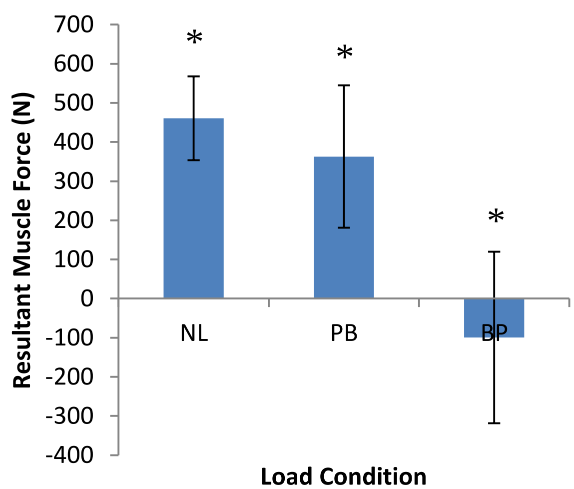


Figure 4.15: Estimated resultant muscle force by load condition. * At α = 0.05, differences were seen between all conditions.

The resultant muscle force averages, seen in Figure 4.15, were statistically different between all pairs of conditions, increasing in magnitude from backpacking to piggybacking to no load. The backpack condition had a negative value and the other two conditions had a positive value; the trunk extensors dominated during backpacking and the trunk flexors dominated during both piggybacking and unloaded walking.

The estimated compressive, shear and resultant joint reaction force values are illustrated separately in Figures 4.16, 4.17, and 4.18 respectively and together in Figure 4.19 for comparison. The compressive and resultant joint reaction force values were both statistically different between each of the three conditions, increasing from no load to piggybacking to backpacking. The estimated shear joint reaction force value increased from no load to each of the identical load condition values. This equality was a direct and predictable result of the lumbar spine model; the shear calculation depended on total load, for which the loaded conditions had identical values.

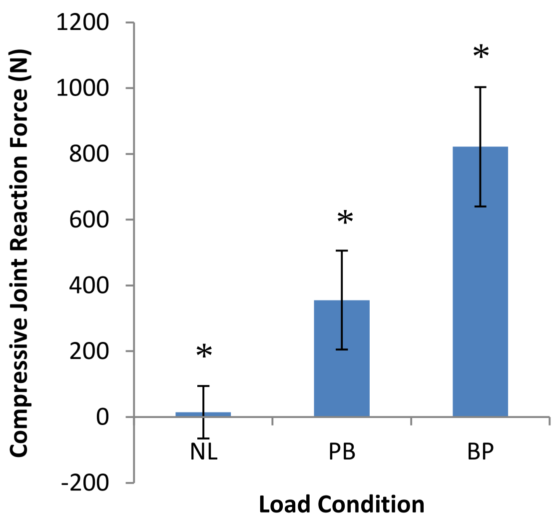


Figure 4.16: Estimated compressive joint reaction force by load condition. *At $\alpha = 0.05$, a difference was seen between all three conditions.

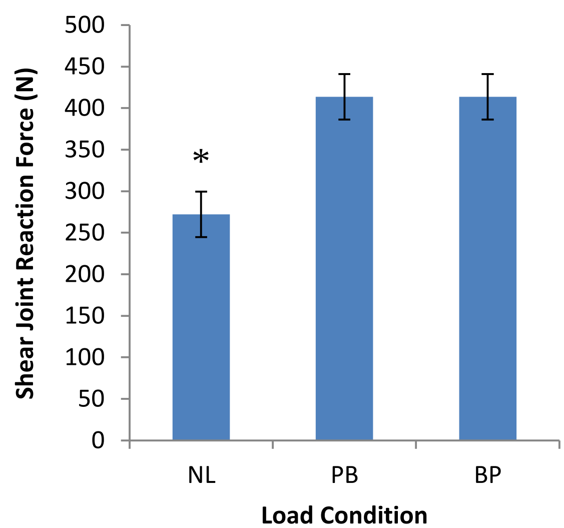


Figure 4.17: Estimated shear joint reaction force by load condition. *At $\alpha = 0.05$, the shear joint reaction force was less for the no load condition.

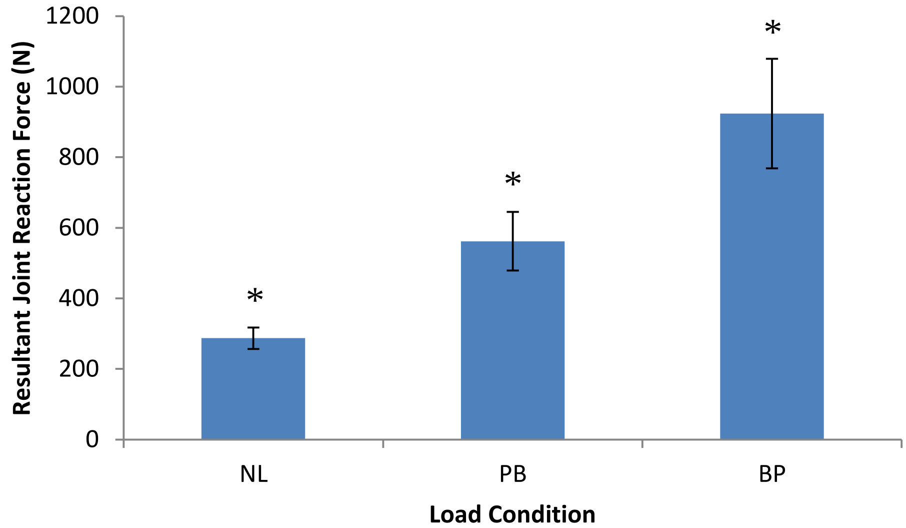


Figure 4.18: Estimated resultant joint reaction force by load condition. * At α = 0.05, the resultant joint reaction force was different between all load conditions.

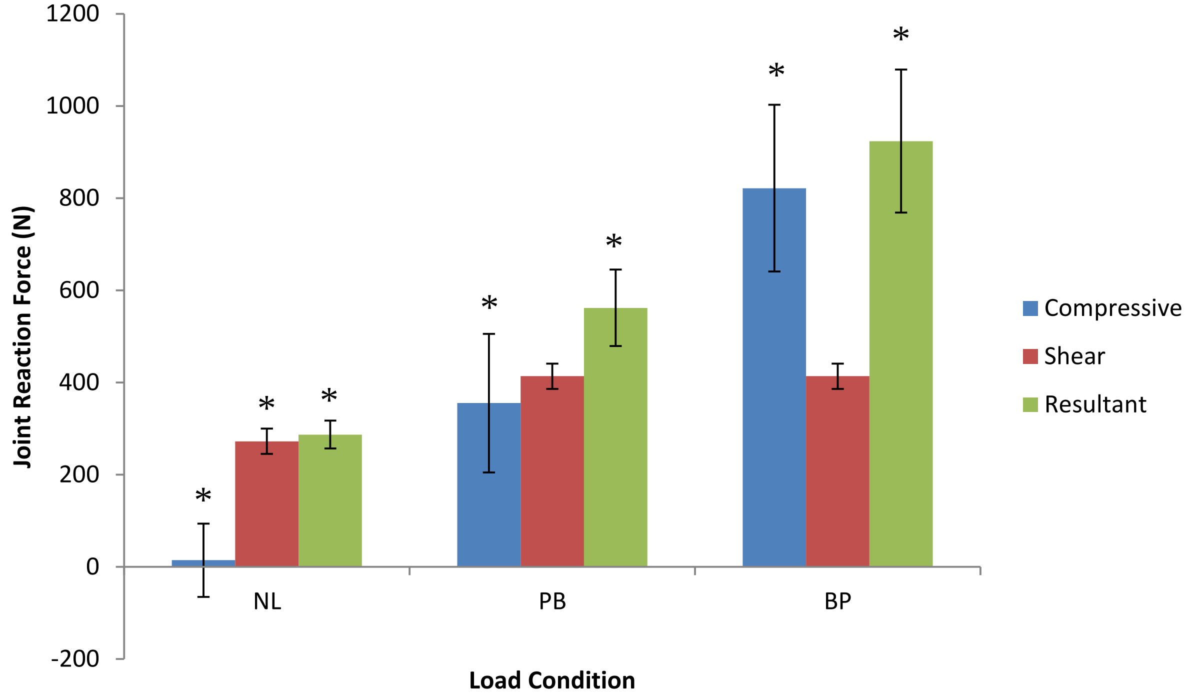


Figure 4.19: Estimated shear, compressive, and resultant joint reaction forces by load condition. * At α = 0.05, the compressive and resultant joint reaction forces were different between all load conditions; the shear joint reaction force was less for the no load condition than for the loaded conditions.

VERTICAL GROUND REACTION FORCE VARIABLES

Walking over a force plate usually generates a predictable butterfly pattern (Robertson et al., 2004), which was primarily seen here. The curves for each trial were slightly different from each other: some with smoother and less variable lines than others, some first peaks greater than the second peaks, and some the opposite. The mean first peak value was less than the mean second peak value when walking unloaded and when backpacking, whereas the peaks were almost equal when piggybacking. The curves tended to be more similar within participants and across conditions than between participants, suggesting that each individual has a specific way of moving that they will repeat when they are asked to repeat an activity. An example of the vertical ground reaction force curves created by each participant is in Figure 4.20. This curve represents the average of three trials by one participant while piggybacking.

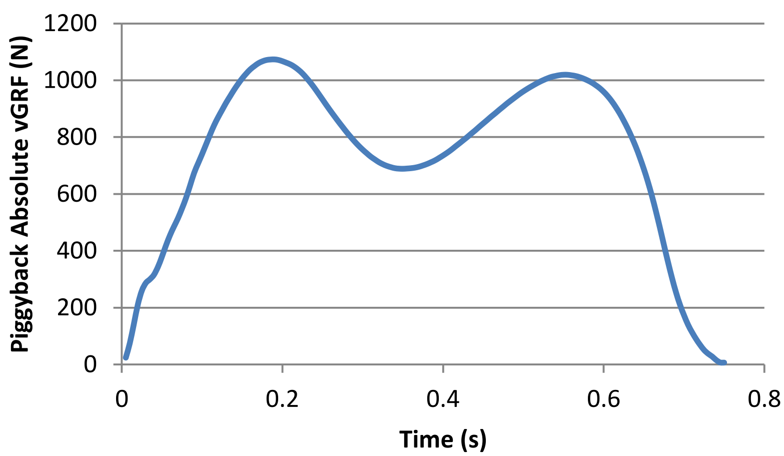


Figure 4.20: A typical vertical ground reaction force curve for one participant, representing absolute vGRF for three averaged piggyback trials.

Relative vertical ground reaction force variable results are listed in Table 4.4 and are illustrated in Figure 4.21. The first peak was only different between piggybacking and backpacking, the interpeak minimum exhibited no differences, and the second peak was only different between no load and piggybacking. For all conditions, the mean first vGRF peak was measured at slightly greater than body weight, increasing from backpacking to no load to piggybacking. All interpeak minimums were approximately three quarters of body weight. The second vGRF peak was greatest for the unloaded condition, less for the backpack condition, and

Table 4.4: Relative vertical GRF variable means and standard deviations by total weight (*TW), with pairwise p-values across conditions.

Variable\Condition	No Load	Piggyback	Backpack	Bonferroni p-value NL/BP	NL/PB	BP/PB
1st Peak (*TW)	1.08±0.09	1.09±0.06	1.05±0.06	.305	1.000	**.011**
Interpeak Minimum (*TW)	0.75±0.06	0.72±0.06	0.72±0.08	.102	.073	1.000
2nd Peak (*TW)	1.12±0.05	1.06±0.04	1.09±0.05	.095	**.001**	.119

- A Bonferroni post hoc test was used when the repeated measures ANOVA determined significance
- Statistically significant p-values are bolded, α = 0.05

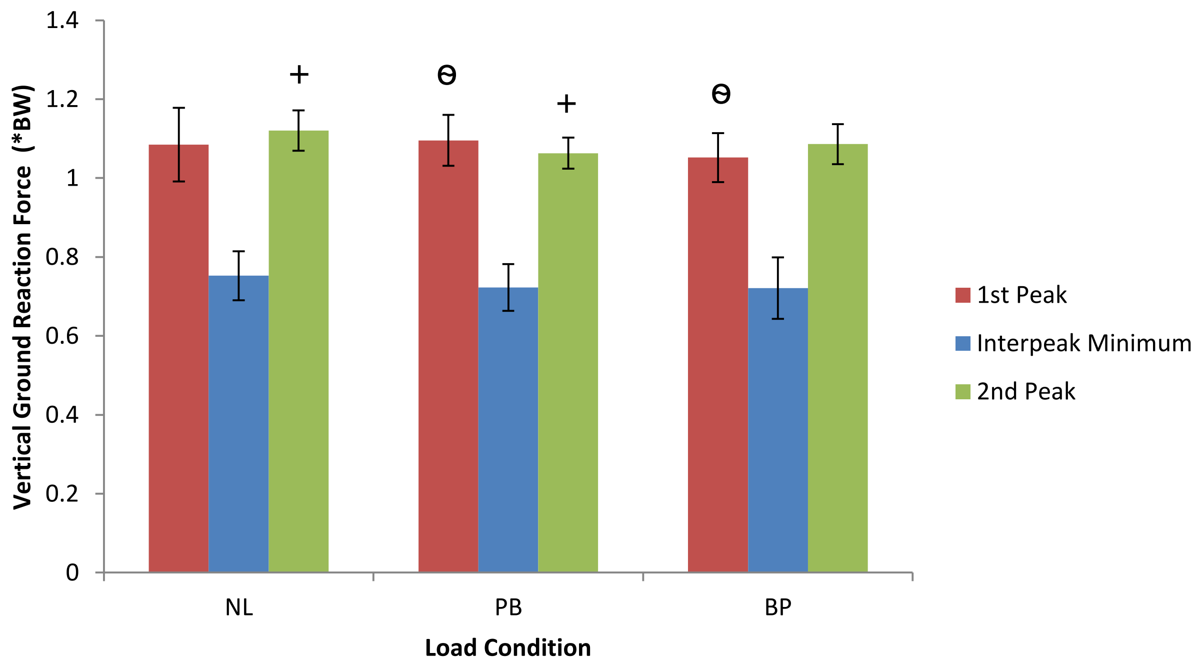


Figure 4.21: The relative 1st peak, interpeak minimum, and 2nd peak of the vertical ground reaction force curves for each load condition. The arrangement of these three variables represents the butterfly shape recorded for each condition. At α = 0.05, the first peak varied between loaded conditions (ϴ) and the second peak varied between no load and piggybacking (+).

the least for the piggyback condition. Participants exerted a greater second relative vertical ground reaction force peak when carrying no load as compared to piggybacking. Therefore, the participants exerted a smaller downward force during the propulsive phase of piggybacking than during the propulsive phase of the no load condition.

Table 4.5: Absolute vertical ground reaction force variable means, standard deviations, and pairwise p-values across conditions.

Variable\Condition	No Load	Piggyback	Backpack	Bonferroni p-value		
				NL/BP	NL/PB	BP/PB
1st Peak (N)	875.75±107.85	1196.62±106.04	1059.84±345.70	**.000**	**.000**	**.013**
Interpeak Minimum (N)	608.74±82.52	790.06±86.61	788.50±107.84	**.000**	**.000**	1.000
2nd Peak (N)	904.99±94.73	1161.98±100.58	1186.64±99.33	**.000**	**.000**	.109

- A Bonferroni post hoc test was used if the repeated measures ANOVA determined significance
- Statistically significant p-values are bolded, α = 0.05

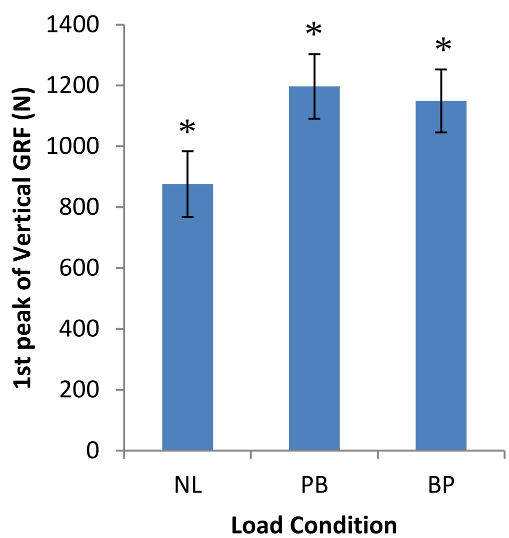


Figure 4.22: Absolute values of the 1st peak of the vertical ground reaction force curve by condition. *At α = 0.05, the 1st vertical ground reaction force peak was statistically different between all conditions.

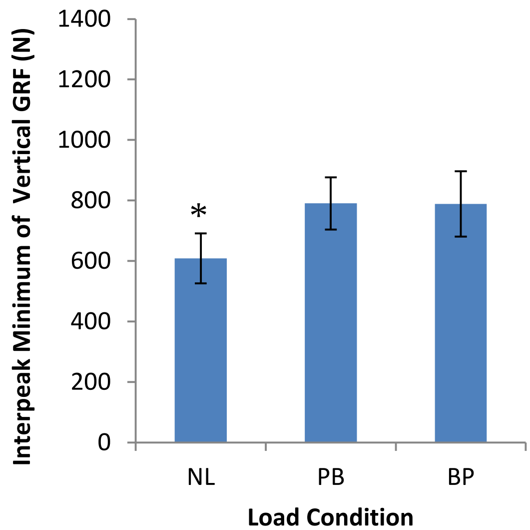


Figure 4.23: Absolute values for interpeak minimum of the vertical ground reaction force curve by condition. *At α = 0.05, the interpeak minimum was less for no load than each of the loaded conditions.

Absolute vertical ground reaction force variable results are listed in Table 4.5 and are illustrated in Figures 4.22, 4.23, 4.24, 4.25. The no load condition produced a lower vGRF than each loaded condition during the first peak, the interpeak minimum, and the second peak of the vGRF curve. The only difference between the loaded conditions was that the first absolute vGRF peak was less for piggybacking than backpacking.

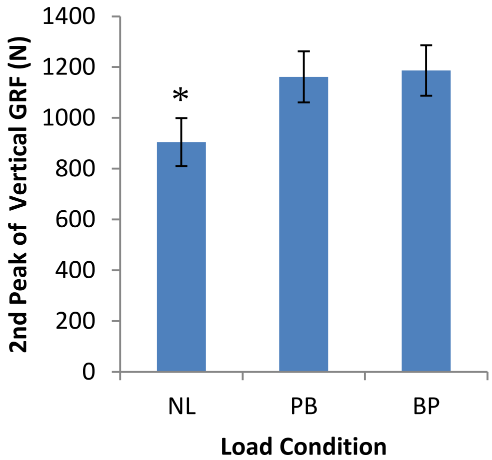


Figure 4.24: Absolute values of the 2nd peak of the vertical ground reaction force curve by condition. *At $\alpha = 0.05$, the 1st vertical ground reaction force peak was statistically less during the no load condition than each of the loaded conditions.

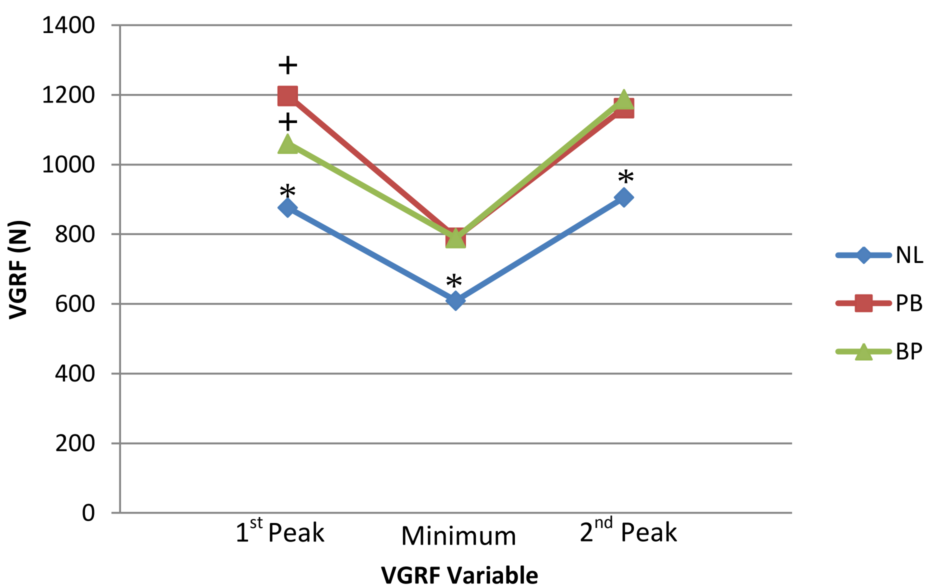


Figure 4.25: The absolute 1st peak, interpeak minimum, and 2nd peak of the vertical ground reaction force curves for each load condition, illustrating the typical butterfly patterns. *At $\alpha = 0.05$, the no load condition was less than each of the two loaded conditions for every absolute vGRF variable. The first vGRF peak value was also less for piggybacking compared to backpacking (+).

Table 4.6: Amount of vGRF produced by the added 284.49 N load, as piggyback passenger or backpack.

	PB		BP	
	N+	%φ	N+	%φ
First Peak	320.88±81.65	112.79±28.70	273.40±52.98	96.10±18.62
Interpeak Minimum	181.31±37.83	63.73±13.30	179.75±51.68	63.18±18.17
Second Peak	256.99±30.48	90.33±10.72	281.65±45.92	99.00±16.14

+ Value represents difference between load condition and no load condition forces at that instant
φ Value represents the N force value as a percentage of 284.49 N.

Comparisons were also made between the vertical ground reaction force differences from each of the loaded conditions to the no load condition, as seen in Table 4.6. The piggyback condition demonstrated a greater difference from the no load condition for the first vGRF peak than did the backpack condition, as seen in Figure 4.26. As a result of adding the 284.5 N load, the mean vGRF increased by 320.88 N (112.79% of the load weight) when piggybacking and 273.40 N (96.10% of the load weight) when backpacking. The first peak paired t-test had a p-value of 0.004, which demonstrated that piggybacking allowed proportionally more impact with the ground. Piggybacking caused a greater first peak vGRF than no load (112.79±28.70%), but backpacking caused less first peak vGRF than no load (96.10±18.62%).

The piggyback and backpack vGRF interpeak minimums increased by approximately the same amount (180 N or 63% of the load weight) from the no load vGRF interpeak minimums. These values are seen in Figure 4.27, and are not significantly different.

At the second peak of the curve, a greater percentage of the load was translated into vGRF for backpacking (282 N or 99%) than for piggybacking (257 N or 90%). This comparison is illustrated in Figure 4.28 and the paired t-test had a p-value of 0.03.

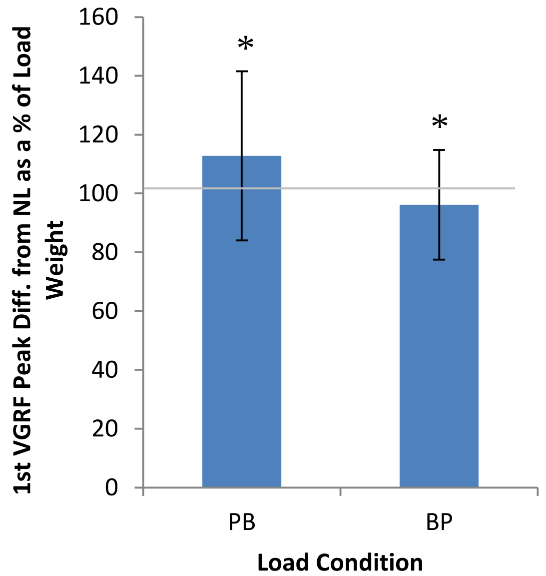


Figure 4.26: The percentage of the external load that demonstrated as additional 1st vGRF peak as compared to the no load condition. *At α = 0.05, the % of load that demonstrated at the first vGRF peak was greater for piggybacking than backpacking.

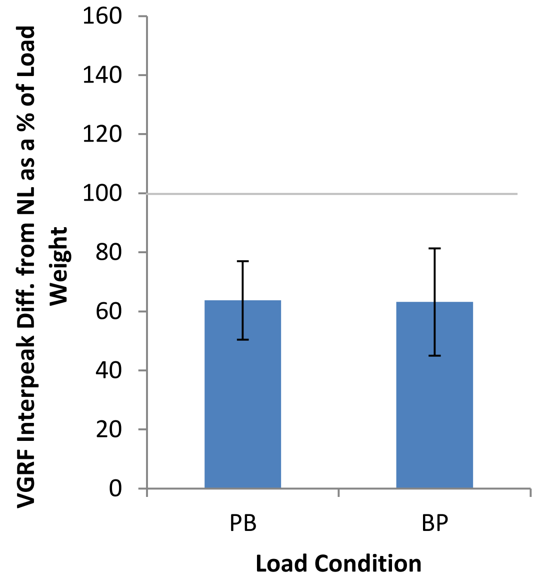


Figure 4.27: The percentage of the external load that demonstrated as additional vGRF interpeak minimum as compared to the no load condition. *At α = 0.05, there was no difference in the % of load demonstrated at the vGRF interpeak minimum across conditions.

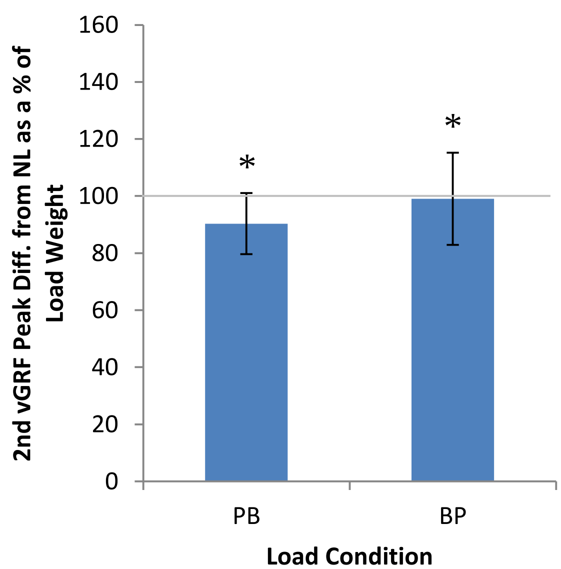


Figure 4.28: The percentage of the external load that demonstrated as additional 2nd vGRF peak as compared to the no load condition. *At α = 0.05, a greater percentage of the load weight translated into the 2nd vGRF peak during backpacking than during piggybacking.

GAIT VARIABLES

Table 4.7: Gait variable means, standard deviations and pairwise p-values across conditions

Variable\Condition	No Load	Piggyback	Backpack	Bonferroni p-value		
				NL/BP	NL/PB	BP/PB
Walking Velocity (m/s)	1.29±0.13	1.26±0.06	1.25±0.13	.708	.677	1.000
Walking Cadence (steps/min)	104.30±5.14	103.41±4.50	101.82±4.96	.286	1.00	.198
Average Step Length (m)	0.75±0.06	0.75±0.05	0.75±0.05	1.000	1.000	1.000
Right to Right Stride Length (m)	1.45±0.11	1.51±0.12	1.48±0.11	.645	.059	.210
Left to Left Stride Length (m)	1.53±0.12	1.50±0.08	1.51±0.10	1.000	1.000	.733
Right Foot Stance Time (s) – FP	0.714±0.040	0.747±0.034	0.766±0.037	**.000**	**.023**	**.008**
Single Stance Time (s)	0.42±0.04	0.40±0.03	0.40±0.02	.512	.429	1.000
Double Stance Time (s)	0.17±0.02	0.19±0.02	0.19±0.02	**.001**	.358	.660

- A Bonferroni post hoc test was used if the repeated measures ANOVA determined significance
- Statistically significant p-values are bolded, $\alpha = 0.05$

The results for all of the gait variables are presented in Table 4.7. There were no differences between any of the conditions for the following variables: walking velocity, walking cadence, average step length, right to right stride length, left to left stride length, and single stance time. Walking velocity and walking cadence results are seen in Figures 4.29 and 4.30, and were not different between conditions.

Right foot stance time was shortest with no load, slightly longer when piggybacking, and longest when backpacking. Right foot stance time is illustrated in Figure 4.31. Double stance time, which was measured with a less accurate method than right foot stance time, only demonstrated a difference between the no load and backpack conditions, with a longer double stance time when carrying a backpack. Double stance time is graphed in Figure 4.32.

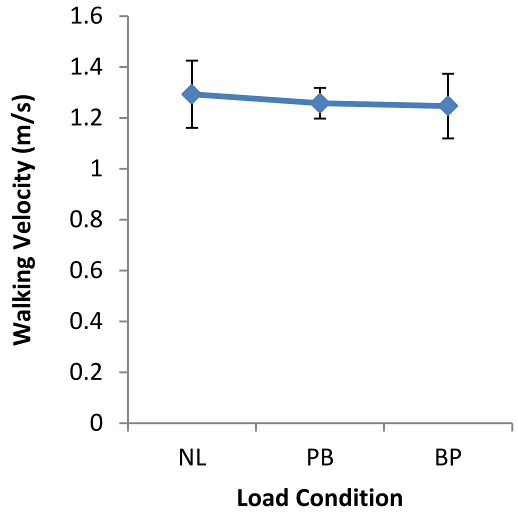


Figure 4.29: Walking velocity by load condition. No differences at $\alpha = 0.05$.

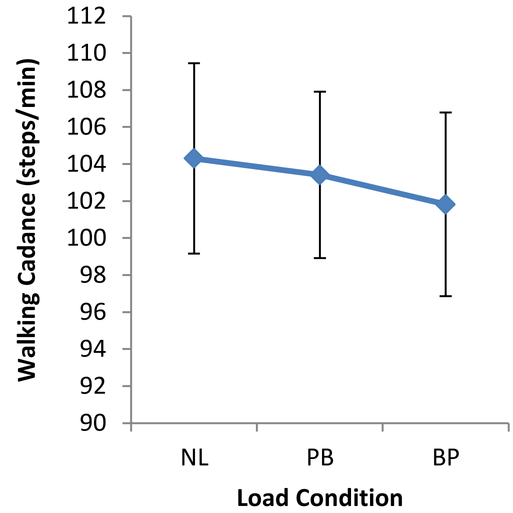


Figure 4.30: Walking cadence by load condition. No differences at $\alpha = 0.05$.

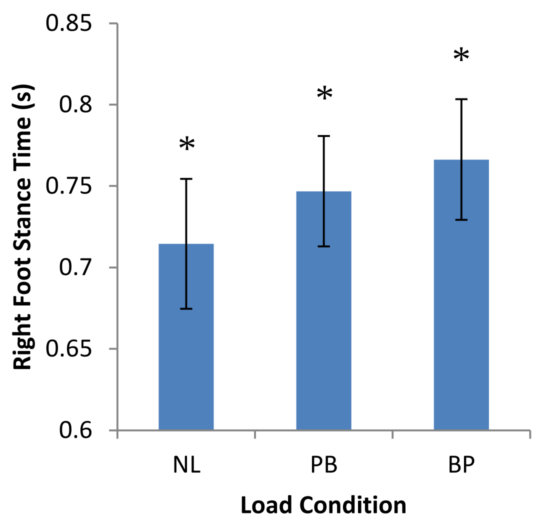


Figure 4.31: Right foot stance time by load condition, as measured by a force plate at 200 Hz. *At $\alpha = 0.05$, the right foot stance time increased from no load to piggybacking to. backpacking.

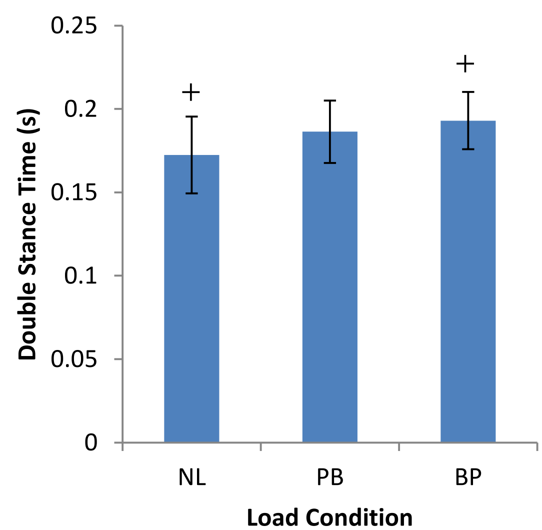


Figure 4.32: Double stance time by load condition, as measured using Dartfish Software at 30 Hz. $^{+}$At $\alpha = 0.05$, the double stance time increased significantly only from no load to backpacking.

CHAPTER V

DISCUSSION

KINEMATIC VARIABLES

Maximum Trunk Flexion Angle

The addition of a load primarily posterior to the trunk caused trunk flexion, which was supported by several previous studies (Gillet et al., 2006; Goh et al., 1998; Kinoshita, 1985; Rodriquez-Soto et al., 2003; Vacheron et al., 1999). This flexion also increased between equal loads from piggybacking to backpacking. The mean trunk flexion angle values were 0.9±1.1°, 9.7±3.1°, and 12.4±2.3° for walking with no load, piggybacking, and backpacking, increasing from no load by approximately 11.5° with the addition of the 29 kg backpack. Previously, the addition of a 15 kg backpack increased trunk flexion during gait from 4.9±2.7° to 14.2±3.2° (Gillet et al., 2006). That load was lighter than the current 29 kg load, and the trunk flexion increase was slightly less. When 3 kg of that load was shifted anteriorly, similar to piggybacking, the trunk angle was 10.6±2.9° (Gillet et al., 2006). This front-back pack trunk flexion angle was between the angles for no load and fully posterior conditions, similar to the current piggyback values. Other researchers measured a 25° to 34° increase in trunk flexion with a 58.8 kg backpack and body armor load (Rodriguez- Soto et al., 2013); this greater load demonstrated greater trunk flexion than in the present study, which is reasonable. There is also likely variation between studies accounted for by differences in landmarks used, researcher methods, and participant populations, for example.

The present researcher's original hypothesis for the cause of trunk flexion with a posterior load was supported by an article by Goh et al. (1998) who first articulated the term "compensatory trunk flexion". This term (Goh et al., 1998) suggested that participants instinctively shifted their body segments forward to create forward torque to counteract the backwards torque created by an added posterior load. Gillet et al. (2006) suggested that the backwards rotating moment caused by the posterior load is compensated for by an increase of six to seven degrees of trunk flexion. The results of the present study also support this compensatory concept, illustrated in Figure 2.20 on page 76. The loads added posterior to the trunk shifted the centre of gravity of the system posteriorly, creating an extensor torque which would likely cause the participant to fall backward if not for the reactive trunk flexion that was exhibited (Goh et al., 1998). The further posterior the centre of gravity or the greater the mass of the load, the greater the posterior shift in the centre of gravity of the system, and the greater the trunk flexion to be expected. The trunk flexes to shift the centre of gravity anteriorly so that it is over the base of support (McGill, Marshall, & Anderson, 2013).

Although there is no previous piggyback loading data to compare present results to, it is evident that the passenger caused similar reactionary trunk flexion to backpacking, but not as great. The increase in trunk flexion from no load to both loaded conditions was due to the addition of 29 kilograms of external load, whereas the increase in trunk flexion from piggybacking to backpacking was due only to the change in method or position of load carriage. The location of the centre of gravity of the load was more posterior for the backpack than the passenger; the backpack load was considered entirely posterior to the lumbosacral joint while part of the passenger load was anterior and/or lateral to the lumbosacral joint. Because the piggyback condition added some anterior and lateral load in comparison to the purely posterior load of the backpack, it did not result in as great a posterior shift in the system centre of gravity when the load was added, nor as great of trunk flexion to shift the centre of gravity back to near its original position over the base of support. Therefore, this difference in load centre of gravity position resulted in different torques about the lumbar spine and ultimately resulted in a greater increase in trunk flexion in order to better balance the system during backpacking. This supports the idea that the position of the load affects gait biomechanics, and illustrates that trunk flexion deviates further from natural during backpacking than during piggybacking gait.

The conclusion that piggybacking allows for a more natural gait than backpacking is not only supported by the results of several variables in the present study, but also by load carriage studies that involved front-back packs (Gillet et al., 2006; Kinoshita, 1985; Legg, 1985). The front portion or pocket of a front-back pack is comparable to the arms and legs of a piggyback passenger, as they wrap around to the front of the body. Therefore, after producing the results of the current study, it remains that front-back packs are the closest representative to piggybacking in existing literature, and that they exhibit less deviation from normal gait when compared to backpacking.

As discussed in Chapter II, trunk flexion increases the stress on the intervertebral facet and disc joints. Trunk flexion causes compression of the anterior disc and tension on the posterior disc, which are resisted by the layers of the annulus fibrosus; this places much greater stress on the disc than rotation because the annulus fibrosus fibres are designed to better resist rotation (Hall, 2007; Levangie & Norkin, 2005; Nordin & Frankel, 2012). As the anterior disc is compressed and posterior disc is in tension, trunk flexion causes the nucleus pulposus to be pushed posteriorly into the tensed and therefore weakened posterior annulus. The narrow posterior longitudinal ligament at the posterior disc does not provide substantial support (Levangie & Norkin, 2005). When the spine is loaded, all of these forces would increase further. While carrying a 29 kg load, the trunk flexion was not great and therefore the discs would not be expected to be placed at substantial risk. However, greater load causes greater trunk flexion so there may be a point when the discs do start to experience dangerous levels of the above-mentioned stress; this load limit would be less for backpacking than for piggybacking because equal loads cause less trunk flexion for piggybacking.

Although it could not be seen through the backpack and passenger, the lumbar lordosis angle probably decreased with the increase in trunk flexion caused by a heavy backpack load (Hall, 2007). Rodriquez-Soto et al. (2013) also used magnetic resonance imaging and saw this effect.

Minimum Trunk Flexion Angle

The angle of minimum trunk flexion during gait was greater when carrying a load as compared to carrying no load. With no load, the trunk was approximately vertical (-0.1±1.3°), whereas with either 29 kg load (PB: 7.3±3.2°; BP: 8.2±3.2°), the trunk exhibited three to twelve degrees of flexion when most upright. This suggested that not only does the maximum angle of trunk flexion increase with load, the trunk of most adult males usually does not reach a vertical position when walking with a load. Participants tended to flex their trunk forward into a "balanced position" and did not excessively deviate from this position during gait. Their trunks did not reach a vertical position because the centre of gravity of the system would have moved posterior to the base of support, which would have resulted in falling backwards. The participants walked while holding their trunk and the load in this constantly flexed position. The amount of this trunk flexion sometimes changed over the gait cycle, however, that depended on the position of the load.

Trunk Flexion-Extension Range of Motion

The range of trunk flexion-extension motion was similar for no load and piggybacking (NL: 2.2±1.1°; PB: 2.3±1.1°), but greater for backpacking (4.2±1.7°). This corresponded to studies suggesting that the trunk ranges only a few degrees during unloaded gait (Goh et al., 1998; Cappozzo, 1981; Thorstensson et al., 1984; Winter et al., 1993), but varied from the results of the following studies. Vacheron et al. (1999) found trunk range of motion during gait to decrease with the addition of a 22.5 kg backpack and Goh et al. (1998) found that trunk range of motion was usually approximately five degrees despite carrying no load, or a 15% or 30% body weight backpack. Vacheron et al. (1999) found a decreased range of motion for the S1-L3-T12 segment of the spine specifically, as opposed to the overall spine, so the variable was slightly different. Participant differences may explain the results by Goh et al. (1998).

Because the minimum trunk flexion angle increased from carrying no load to carrying either load, but did not differ between loaded conditions, the difference in the range in trunk range of motion between piggybacking and backpacking resulted from the difference in maximum angle of trunk flexion. This is illustrated in Figure 4.3 on page 123 of Chapter IV. While the participants often returned to nearly the same least flexed position when piggybacking and backpacking, they ranged from that position to their maximal flexion position, which was greater for backpacking.

Backpacking exhibited approximately twice the amount of trunk range of motion as compared to the other two conditions; the masses of each segment would have needed to be accelerated twice as fast each gait cycle to increase and decrease trunk angular velocity. This increase in the range of motion when walking would not only necessitate more stabilization and muscle coordination to control the trunk, but would also increase the forces experienced at the lumbosacral joint. The lumbosacral spine model assumed equilibrium when calculating the muscle torque, muscle force, and the joint reaction forces. Therefore, these forces would be greater if accounting for accelerations of the trunk; a greater trunk range of motion would suggest greater forces. Accepting that the static model accounts for zero acceleration and provides an underestimation of the lumbosacral forces, it is important to understand that the greater the actual range of trunk motion during gait, the more the actual forces are underestimated by the model. Backpacking is an example of this.

Trunk flexion, along with hip and knee flexion, are actions in the sagittal plane. It has been suggested that loaded gait has more sagittal plane motion and less transverse plane motion as compared to unloaded gait (Atreya et al., 2010). This is likely because transverse plane rotations would demand additional muscle control in the transverse plane in order to control the accelerations of the mass of the load at its distance from the longitudinal axis of rotation (Atreya et al., 2010). The centre of gravity of the backpack was at a distance posterior to the longitudinal lumbosacral axis of rotation and would therefore require a good deal of acceleration and deceleration. Although transverse plane motion was not measured in the present study and so any alterations caused by loading cannot be confirmed, the sagittal plane trunk range of motion did increase, which supports this theory.

The trunk range of motion variable supported that piggybacking gait is more similar to natural gait than is backpacking gait. Because angular oscillations of the trunk are one of the most common reasons thought to increase symptoms of back pain associated with carrying a load (Vacheron, 1999), this may suggest a greater risk of back pain with backpacking than with piggybacking.

Maximal Hip and Knee Flexion Angles

The maximal angle of hip flexion increased from no load to each loaded condition. However, this is believed to be a reflection of the increase in trunk flexion angle with load because the hip measurement was taken from the line that extends from the trunk. As this reference line changed in orientation, so did the measured relative hip angle. When an absolute hip flexion angle was considered by subtracting the trunk flexion angle from the hip flexion angle, there were no differences between conditions. An increase in sagittal plane motion was expected with the addition of load (Birrell et al., 2005; Chow et al., 2005); this was not found with respect to hip flexion angle but the result was different for knee flexion angle.

The angle of maximal knee flexion increased from no load (17.8±8.9°) to backpacking (24.2±3.7°) and from piggybacking (22.6±3.0°) to backpacking. With the addition of a load, the carriers may have subconsciously altered gait to lessen the increased impact with the ground (Chow et al., 2005). This "cushioning" could be performed by any combination of joints in the sagittal plane. Because the absolute hip angle did not alter and the knee flexion angle did increase when backpacking, it may be suggested that this cushioning was primarily performed by the knee joint. The ankle may also contribute cushioning effects, but this has not yet been investigated.

Because there was an additional 29 kg of mass to be controlled and absorbed at impact with the ground, the piggybacking knee flexion angle would be expected to be greater than the no load knee flexion angle even if other gait mechanics were identical. The knee flexion results support the theory that piggybacking is biomechanically more similar to natural gait than is backpacking.

EXTERNAL LUMBOSACRAL TORQUE VARIABLES – LOAD TORQUES

Every external lumbosacral torque variable was different between all three conditions except for the total amount of negative torque.

Resultant Torque Produced by Load Only

The resultant torque caused by the passenger and backpack were both positive or backward, but the passenger (16.68±4.81 Nm) caused less backward torque than the backpack (54.69±6.67 Nm) because of their different positions and resulting moment arms. The moment arm from the centre of gravity of the backpack to the lumbosacral joint was longer than the sum of the moment arms from the centre of gravity of each passenger segment to the lumbosacral joint; the weight of the backpack load acted at a greater distance from the axis of rotation than did the overall weight of the passenger load. This longer moment arm for an equal load resulted in a greater positive torque produced by the backpack than the passenger (Hall, 2007).

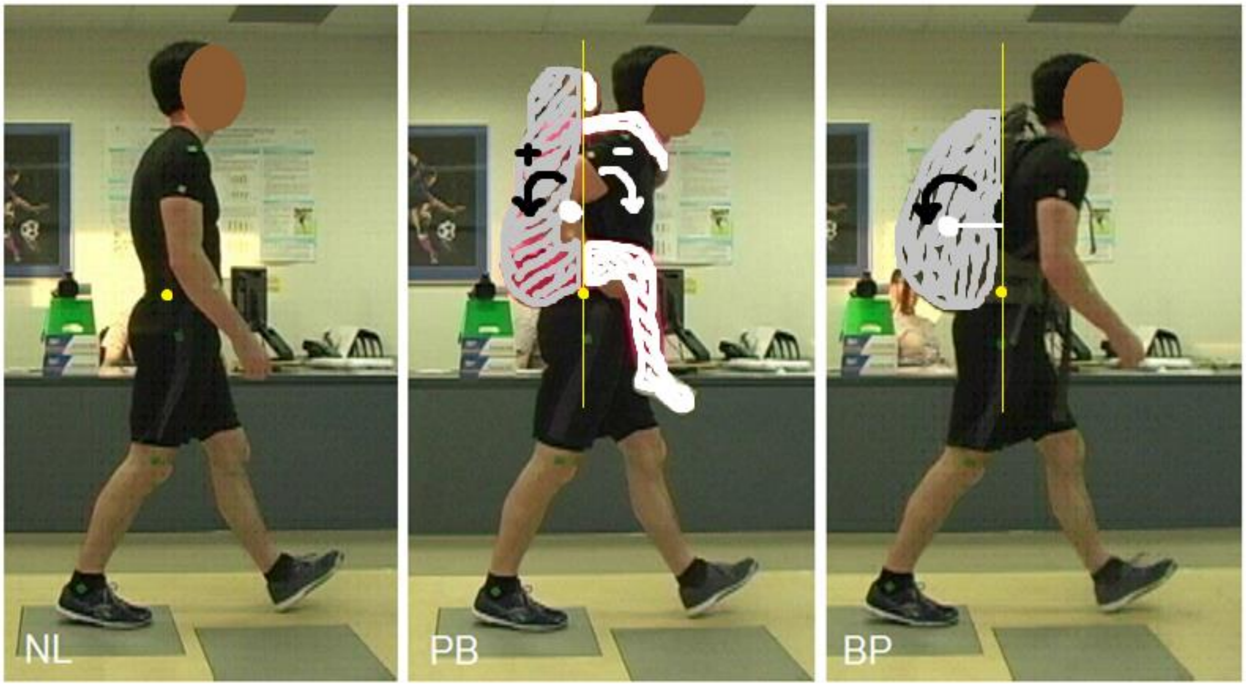


Figure 5.1: A comparison of positive and negative torques produced by the load and the position of the overall centre of gravity of the load with corresponding moment arms. Backward-rotating torques are represented by grey with black arrows and forward-rotating torques are represented by white and a white arrow. The centre of gravity of each load is represented by a large white dot. The moment arm of each load is represented by a white horizontal line connecting its centre of gravity to the vertical line at the lumbosacral joint.

Several segments of the passenger load were anterior to the lumbosacral joint with a moment arm extending anteriorly from the joint rather than posteriorly, which indicated the production of negative rather than positive torque. The positive and negative torques produced by the centre of gravity of each load are illustrated in Figure 5.1. The passenger caused approximately one quarter negative torque and three quarters positive torque. The existence of this negative torque decreased the resultant positive torque caused by the passenger. Overall, the external load effects were different for the passenger, which produced 25.13±3.88 Nm backward and 8.45±1.51 Nm forward, and the backpack, which only produced 54.69±6.67 Nm of backward torque. This is illustrated in Figure 5.2.

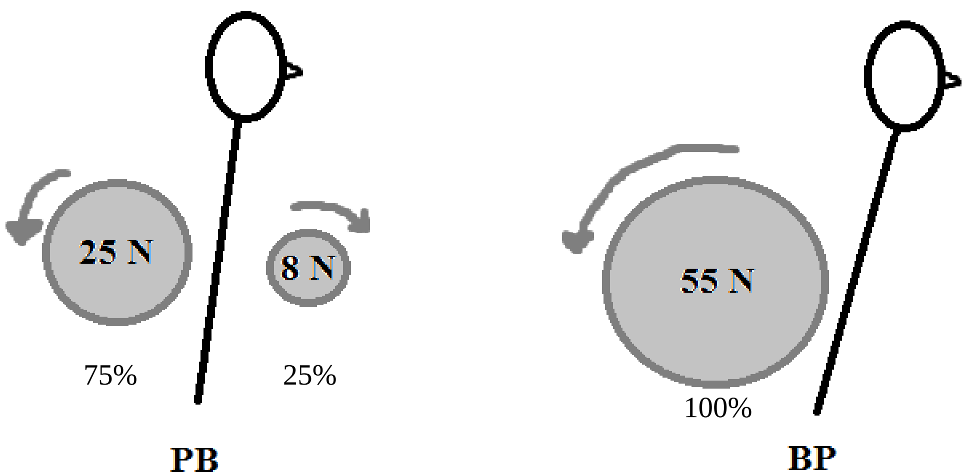


Figure 5.2: A comparison of the forward and backward mediolateral lumbosacral joint torques produced during piggybacking and backpacking by the 29 kg load alone. The positive and negative magnitudes are subtracted to determine resultant load torque and added to determine total magnitude of load torque.

Backpack and piggyback loads both have an overall backward rotating tendency, but a backpack load has a greater tendency to produce backward rotation than does an equal piggyback load. This stronger tendency for rotation has substantial effects on the position of the carrier as compensation for the resultant torque, demonstrated by the increased angle of trunk flexion, deeper knee flexion and longer right foot stance time found in the present study. Piggyback loads also acted to pull the trunk into backward rotation but did not create as great of a backward torque and unbalance and therefore produced less deviation from natural gait. This was demonstrated by similar but lesser adaptations in trunk flexion, knee flexion and right foot stance time. Because of this, piggybacking may be considered a more natural method of load carriage.

Total Magnitude of Torque Produced by Load Only

When the magnitudes of the positive and negative portions of the load torque were summed to the total magnitude of load torque, the passenger (33.42±3.33 Nm) caused less total magnitude of torque than the backpack (54.69±6.67 Nm) even though the weights were equal. This was due to the longer moment arm of the backpack weight in comparison to the sum of the moment arms for the passenger segment weights as described above. The body segments of the passenger tended to be closer to the lumbosacral joint than the backpack, with some in positive and some in negative directions. The segments that were further from the joint and were in the position to create larger torques only had small masses such as the 8.36 N feet. The greater magnitude of torque produced by the backpack suggests that backpacking is generally more strenuous than piggybacking.

Torque Produced by Body Segments of Carrier

In the present study, the upper body segments of the carrier were always in a position to produce positive torque at the lumbosacral joint, but the amount of torque varied by condition. When carrying no load, a small amount of torque (-27.66±6.42 Nm) was produced, due only to the posterior position of the lumbosacral joint within the trunk. Therefore, all segment weights were anterior to the joint and in positions to cause forward torque. When piggybacking, the posterior load produced reactionary trunk flexion which moved the centres of gravity of the upper body segments anteriorly, into positions with greater moment arms. Therefore, the same upper body segment weights were shifted to positions that resulted in greater forward torque (-38.70±7.27 Nm). The greatest backward load torque was demonstrated during backpacking, which resulted in the greatest trunk flexion and therefore the longest moment arms for the segment weights. Therefore, the greatest torques produced by the upper body segments of the carrier (-48.73±8.75 Nm) occurred while backpacking.

Total Resultant Torque

The total resultant torque is a very important variable as it is the magnitude of torque that the trunk muscles need to produce in the opposite direction to hold the trunk in position while walking; this magnitude is the value that the lumbar spine model used to calculate joint reaction force.

The resultant load torque and the torque produced by the body segments of the carrier were summed to determine the total resultant torque, which increased from no load (-27.66±6.42) to piggybacking (-22.01±10.73) to backpacking (5.96±13.15). The centre of gravity of the system shifted from the position of the centre of gravity of the carrier towards the position of the centre of gravity of the load when it was added. Because the backpack and piggyback loads had equal masses, the more posterior centre of gravity of the backpack resulted in a more posterior centre of gravity of the system. This resulted in a greater backwards-rotating torque, which is discussed as its components below.

When the resultant torque is divided into its components, positive and/or negative load torque and the negative carrier segment torque, it is clear to see how each total resultant torque is calculated. For the unloaded condition, there was no backward torque and only the positive torque caused by the upper body of the carrier, which was minimal at approximately 28 N. The direction of the resultant torque was predictable for the condition with no added load, due to the known anterior location of the centre of gravity of the head, arms and trunk with respect to the lumbosacral axis of rotation. From this starting point with no load, it is logical that the piggyback condition, which had some anterior weight and some posterior weight, also generated a flexor load torque, and the backpack, which had all posterior weight, had enough posterior load to generate an overall extensor torque.

For piggybacking, approximately three quarters of the passenger load created backward torque, one quarter of the passenger load created forward torque, and the upper body of the carrier produced slightly more forward torque than during the unloaded condition. The negative torque of the upper body of the carrier and the arms and legs of the passenger was greater than the positive torque of the head and trunk of the passenger. Although there was some posterior torque to counteract some of the negative torque, the overall resultant torque was forward. Because some forward torque was counteracted, the resultant magnitude was less than for no load.

The backpack created a backward torque which was almost balanced by the upper body segments of the carrier; the backpack only produced approximately 6 Nm more torque than did the upper body segments. This suggested that the trunk instinctively flexed forward to a fairly balanced position in response to the backpack load, possibly to ease some of the muscle work.

These torques can be placed into context by considering that the L4-L5 moments were 61 Nm to 267 Nm when walking and carrying heavy loads (McGill et al.b, 2009), over 200 Nm for other lifting tasks (McGill & Norman, 1985), and over 900 Nm for extreme lifting (Cholewicki et al., 1991). Considering the much less extreme position of load carriage demonstrated in this study, it is reasonable that the torque value was much less.

Total Magnitude of Torque

Although the magnitudes of total resultant torque increased from backpacking to piggybacking to no load, the total magnitude of torque variable clarifies that the actual amount of torques decreased in that order of conditions. While total resultant torque is useful for determining the overall tendency for rotation, the total magnitude of torque explains that the lumbosacral joint is actually experiencing and supporting a much larger amount of torque during backpacking (103.41±8.33 Nm) in comparison to the other conditions. Piggybacking (72.28±5.58 Nm) also exhibited a much greater amount of torque than unloaded walking (27.66±6.42 Nm). Although the no load condition exhibited the highest total resultant torque, its total magnitude was approximately one third of the piggyback torque and one quarter of the backpack torque. The trunk was usually upright during unloaded walking, and with no added load, it was expected that the total magnitude of torque would have been least for this condition.

Total Negative Torque

The total resultant torque, total magnitude of torque, and the magnitudes of torque in each direction can be used together to understand the direction and magnitude of the forces and torques occurring during these conditions. Every condition exhibited negative torque caused by the upper body segments of the carrier but in differing amounts as discussed previously; the anterior passenger segments also contributed to the negative torque for the piggyback condition. It is logical that the total negative torque increased from no load (-27.66±6.42 Nm) to piggybacking (-47.15±7.62 Nm); this is comparing the negative torque produced by only the body of the carrier with the same plus the arms and legs of the passenger. It is interesting that the backpacking condition (-47.73±8.75 Nm) exhibited the greatest negative torque because it was only caused by the body of the carrier. However, this was due to the increased trunk flexion which resulted in increases in the moment arms and torques for each segment.

Total Negative Torque as a Percentage of Total Magnitude of Torque

All torque produced during natural walking was negative. Approximately 65 percent and 47 percent of the torque produced while piggybacking and backpacking, respectively, were negative. Although the body of the carrier during backpacking caused greater torque than the body of the carrier and parts of the passenger during piggybacking, the large amount of posterior torque still caused greater positive torque.

Total Positive Torque

The amount of positive torque increased from no load to piggybacking to backpacking. There was zero positive torque produced for no load. There were 25.14±3.88 Nm of positive torque produced by posterior passenger body segments during piggybacking and 54.69±6.67 Nm of positive torque produced by the backpack. Because the loads were equal in mass and part of the passenger load was anterior to the lumbosacral joint, it was obvious that the positive torque would be greater for backpacking than piggybacking.

Total Positive Torque as a Percentage of Total Magnitude of Torque

The total magnitude of torque was approximately 0, 35, and 53 percent positive torque for no load, piggybacking, and backpacking respectively. In this model, unloaded walking did not exhibit any backward torque, piggybacking exhibited much less backward than forward torque, and just over half of the torque experienced when backpacking was backward-rotating.

Summary of External Torques

Figure 5.3 is a summary of the torques acting during each condition, based on the averages of various variables. Included variables are the trunk flexion angle, total resultant

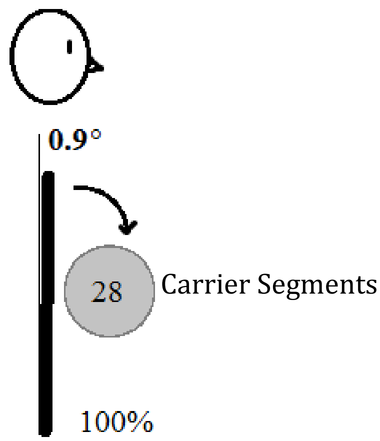
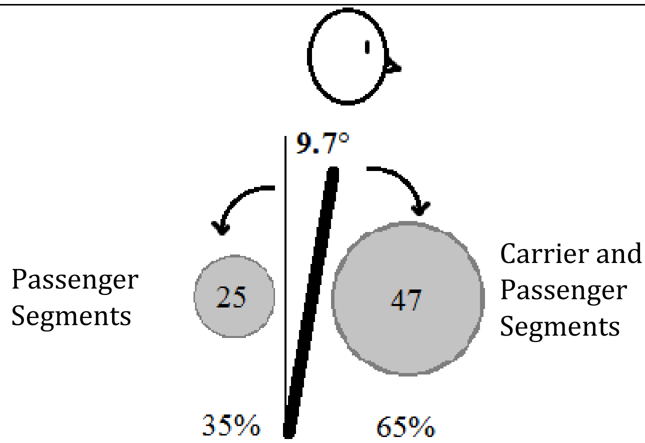
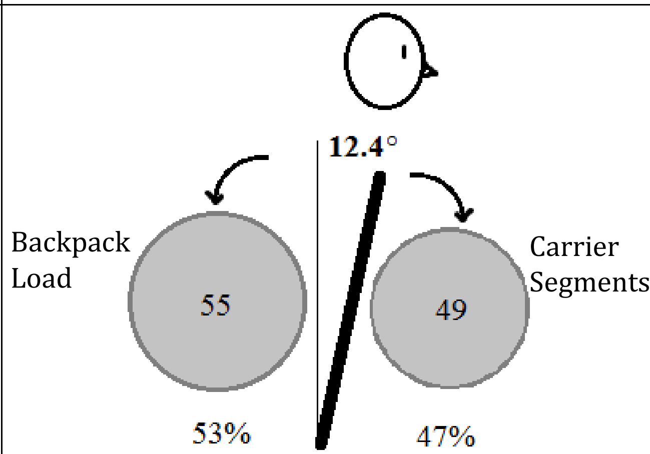
Load	Torques Present (Nm) and Sources of Torque	Rotational Tendency of the Load System	Predominating Muscles based on Resultant Torque
NL		Fall Forward ~ –28 Nm	Erector Spinae
PB		Fall Forward ~ – 22 Nm	Erector Spinae
BP		Fall Backward ~ 6 Nm	Abdominals

Figure 5.3: Schematic representation of the three load conditions, to scale based on averages of trunk flexion angle and positive and negative torque percentages. The grey circles represent load, in which there is a number representing the Nm of torque produced by that portion of the 484.49 N load.

torque, total negative torque, negative torque as a percentage of total magnitude of torque, total positive torque, positive torque as a percentage of total magnitude of torque, and resultant muscle torque. Each variable has been discussed previously except for resultant muscle torque which is found in the next section.

This schematic diagram has been drawn to scale, illustrating comparisons between conditions for the varying trunk angle, and the proportions of anterior and posterior load and torque. Notice that although the only mass causing anterior torque for both no load and backpacking is the upper body of the carrier; the increased trunk flexion angle during backpacking increased the moment arms of the body segments, which resulted in greater torque caused by the same body segments. Both the torque value and percentage of total torque were included to allow these specific findings to be generalized to other loads. The amount of torque caused by the 484.5 N load is represented by grey circles to illustrate the anterior versus posterior torque produced for each condition. This torque as a percentage of the total torque is also listed below the circles to illustrate the proportions of torque that would be experienced while carrying a different magnitude of piggyback or backpack load. The words beside the grey circles describe which masses are producing that particular portion of the load.

This diagram illustrates substantial differences in the torques experienced between these two equally-loaded conditions. Note the similar amounts of positive torque but substantially greater negative torque during backpacking. In both cases, the positive torque was caused by the upper body of the carrier, with the addition of part of the passenger for the piggyback condition. The entire backpack caused negative torque while only the remaining portions of the passenger caused negative torque, and with a lesser moment arm.

Overall, this diagram illustrates that with respect to trunk flexion and torque, the piggyback condition was much more similar to unloaded walking than was the backpack condition. In fact, the no load and piggyback conditions were found to have opposite trunk muscle forces dominating as compared to the backpack condition.

INTERNAL LUMBOSACRAL TORQUE AND FORCE VARIABLES

Resultant Muscle Torque

The resultant muscle torque increased from backpacking (-5.96±13.15 Nm) to piggybacking (21.77±10.91 Nm) to no load (27.66±6.42 Nm); the overall muscle pull rotated the trunk forward during backpacking (trunk flexors) and backward during piggybacking and when unloaded (trunk extensors). This muscle torque was a direct response to counteract the total resultant torque experienced in the opposite direction, in order to maintain trunk position.

According to this model, the trunk flexors produced more tension than the trunk extensors during backpacking, although both groups would have been simultaneously contracting to stabilize the spine (Moore et al., 2010). Al-Khabbaz et al. (2008) chose to measure the activity of the trunk flexors when adding load to a backpack while stationary, which suggests they expected the agonists to be the abdominal muscles. Although they did not find a significant increase in the trunk flexor force, they found increases in trunk flexor tension with added load (Al-Khabbaz et al., 2008).

When piggybacking or carrying no load, both muscle groups would also likely be activated but with the trunk extensors dominating. This supports the theory that piggybacking is biomechanically more similar to unloaded walking than is backpacking. The magnitude of the resultant muscle torque was greater for the no load condition than for the piggyback condition.

The external torque variables explain that this is because although part of the passenger was anterior and the trunk leaned further forward to create greater forward rotating torque in piggybacking, the posterior portions of the passenger still produced greater backward rotating torque which offset the forward torque. The total resultant load torque is therefore less forward rotation than for no load, so the muscle reaction force is less trunk extensor muscle force.

When carrying 29 kg of external mass that needs to be controlled and stabilized during motion, the trunk flexors and extensors would likely exert greater forces than when not carrying an external load. However, the estimated muscle torque was greatest for no load. Therefore, the muscle torque is likely a greater underestimation for backpacking than it is for no load. The piggybacking condition would also have a greater underestimation than no load.

The muscles that flex and extend the trunk were described in Chapter II. The rectus abdominis, external obliques, and internal obliques dominated to flex the trunk (Moore et al., 2010) during backpacking. The erector spinae and multifidi muscles (Hall, 2007; Moore et al., 2010; Nordin & Frankel, 2012) are the primary muscles that produce trunk extension, possibly assisted by the rotatores, interspinalis, and intertransversarii (Hall, 2007; Nordin & Frankel, 2012). As per the results of this study, the above trunk extensors likely dominated when walking with no load and when piggybacking.

The multifidus is responsible for approximately 20 percent of the extensor torque at the lumbosacral level (Bogduk et al., 1992), so it likely produced at least 4.2 Nm and 5.4 Nm of torque for no load and piggybacking. Bogduk et al. (1992) estimated that the lumbosacral extensor muscles produced 144 Nm to 450 Nm of torque while standing. Their participants were either standing differently, their values are overestimations, the present values are underestimations, or a combination of these possibilities.

When considering these results, it is also important to remember that the calculated muscle torque, which all lumbosacral load variables are based on, is a resultant value. It is the difference between the trunk flexor muscle torque and the trunk extensor muscle torque, and the opposite sign and equal magnitude of the total resultant torque.

Resultant Muscle Force

The resultant muscle forces were positive for the no load and piggyback conditions and negative for the backpack condition, corresponding to the resultant muscle torques mentioned above. Therefore, although both the trunk flexors and trunk extensors would have been active to create a stable torso when walking, the trunk flexors dominated by 460.98±107.08 N and 362.80±181.80 N during the no load and piggyback conditions respectively and the trunk extensors dominated by 99.40±219.39 N during the backpack condition. Due to the trunk position demonstrated for each condition, the direction of these muscle forces are logical as discussed in the muscle torque section above. The total force exerted by these muscles is unknown at this time; although an underestimation, this resultant muscle force was the best value available so it was used for the joint reaction force calculations.

For comparison, the mean maximal voluntary extensor muscle forces were approximately 1000 N to 1200 N by the erector spinae muscles, and an extra 614 N by multifidus (Bogduk et al., 1992). As stated above, Al-Khabbaz et al. (2008) documented an increase in abdominal muscle tension with backpack load, but force value differences were not significant. This may have been because the electromyography compares exertions to maximal voluntary contractions and is not necessarily a reliable method of measuring muscle force. Even without considering the weight of the added piggyback or backpack loads, the co-contraction of the muscles stabilizing the spine could have increased the compressive and shear forces by more than the calculated values (McGill et al., 2009a; McGill et al., 2009b).

Lumbosacral Joint Reaction Forces

The magnitude of the lumbosacral forces were affected by the position and amount of load carried (Cholewicki et al., 1991). When the position of the body was altered, as with the increase in trunk flexion, the location of the line of gravity shifted and the forces on the lumbar spine changed accordingly (Levangie & Norkin, 2005).

Lumbosacral Compressive Joint Reaction Force

The compressive joint reaction force would intuitively be greater when carrying a load than when not carrying a load, as was seen during the present study. The compressive lumbosacral joint reaction force increased from 14.49±79.57 N to 355.40±150.27 N to 821.62±181.02 N from no load to piggybacking to backpacking. The compressive force was least for the no load condition, which was expected because the only load supported by the joint was the upper body of the carrier himself. However, the value is very low, appearing equal to the amount of force produced by gravity acting on 1.5 kg of load. There was a standard deviation of 79.57 N, suggesting that there was substantial variation between participants.

It is also logical that the compressive force increased from piggybacking to backpacking because there were substantially greater magnitudes of torque experienced by the backpacking condition. The differences between the conditions were affected by the differences in the muscle forces, as the muscle force was a main part of the compressive joint reaction force calculation. The greater muscle force during backpacking contributed to the greater compressive force experienced at the lumbosacral joint. Because the muscle forces were underestimations, these values are likely underestimations as well, which explains the low value for no load. There was also great variation in these values, with standard deviations of 150.27 N and 181.02 N for piggybacking and backpacking. There certainly would have been variations in gait patterns, but the range may be able to be reduced if the accuracy of the model was improved. An accurate value for muscle force could be found by directly measuring the muscle tension, which would be invasive and risky, rather than calculating the difference between the trunk flexor torque and the trunk extensor torque and converting to force.

These values are much less than the compressive joint reaction force values in the literature. Accepted values for compressive joint reaction forces during walking are in the range of 1.2 to 2.5 times body weight (Cappozzo et al., 1984; Goh et al, 1997; Khoo et al. 1995; Levangie & Norkin, 2005), whereas the lumbosacral compressive forces were 0.02±0.10 times body weight while walking unloaded in the present study. When piggybacking, this rose to 0.33±0.15 times total weight and when backpacking, this rose to 0.76±0.21 times total weight. Although the true values must be substantially greater, this order of increase in force with load is probable (Goh et al., 1997).

The compressive forces during load carriage can also be compared. With a 30% body weight backpack, equal to the lightest load used in the present study, Goh et al. (1997) calculated compressive joint reaction forces of 2.46±0.29 times body weight. Again, this illustrated the extent of underestimation seen here. Goh et al. found that the compressive force increase from unloaded to loaded was 64% (1997), compared to the 1650% for piggybacking and 3800% for backpacking seen here. Other studies also estimated greater values: while carrying only 5kg in each hand, McGill et al. (2013) calculated compressive forces at 1225 N, and actively co-contracting the trunk muscles increased L4-L5 compressive forces by 300 to 1000 N (Kavcic et al., 2004). Normal exercises such as prone arm and leg extensions or lifting 18kg can cause approximately 6000 N of compressive force on the lumbar spine (Levangie & Norkin, 2005; McGill & Norman, 1985). Strenuous lifting can exert over 17 000N of compression (Cholewicki et al., 1991). The maximal permissible limit chosen by NIOSH (1981) is 6377 N. It follows then, that the values calculated here do not represent any risk to the lumbar spine and because they are such large underestimations, their greatest value is to order the conditions according to compressive force.

According to the lumbar spine model, the addition of the 284.49 N passenger resulted in 120% of that load expressed as added compressive joint reaction force while the addition of a 284.49 N backpack resulted in 284% of that load expressed as added compressive joint reaction force. The estimation that piggybacking places less compressive stress on the lumbosacral joint for an equal load compared to backpacking suggests that backpacking may place the lumbar spine at greater risk of injury.

Approximately 80 percent of the axial compression caused by carrying a person or backpack is supported by the intervertebral discs, while 20 to 30 percent (Hall, 2007; Levangie & Norkin, 2005) is supported by the facet joints. Piggybacking and backpacking also alter the internal environment of the discs. Because the intrinsic disc pressure is approximately 150 percent of the external load per unit area, the 484.5 N load would have theoretically increased mean intrinsic pressure to 726.75N per unit area. This pressure would be resisted superiorly and inferiorly by the vertebral endplates and the annulus fibrosus on each side, therefore increasing the stress experienced by all of the structures. The circumferential stress on the annulus would be 1938 N to 2423 N, according to Nordin and Frankel (2012).

Water would have left the discs until the pressure equalized in and out of the disc (Hall, 2007). As the discs decreased in height, the amount of load supported by the facet joints would have increased (Hall, 2007). Once the backpack or passenger was unloaded, the decrease in pressure would have allowed water to flow back into the discs. If a disc is in poor condition, it may not accommodate these forces and changes as easily. Therefore, individuals with previous disc injuries or older individuals should avoid sudden movements while piggybacking or backpacking, especially when first adding the load or when it has just been offloaded.

Lumbosacral Shear Joint Reaction Forces

The shear joint reaction force results were logical, with an increase from no load to loaded. There was 272.24±27.43 N of shear joint reaction force with no load and 413.59±27.43 N when carrying 284.5 N of load on the back. The shear joint reaction force for the two loaded conditions were equal because, with the same weight load carried, there were no real differences in the values entered into the formula. For piggybacking, each of the segments of the passenger load were added separately, but their sum was equal to the mass of the backpack and the calculations therefore had the same result.

The estimated shear joint reaction forces were much closer than the compressive forces to values in the literature. The mean shear forces were approximately 0.34 times body weight and 0.38 times total weight when walking unloaded and with a load, in comparison to 0.22 times body weight and a range from 0.06 to 0.63 times body weight when walking unloaded (Khoo et al., 2005). McGill et al. (2013) estimated that 167±113 N of shear were experienced at L4-L5 when carrying 5 kg in each hand, which would suggest that the present values may be slightly high but the L4-L5 shear joint forces are also expected to be lower than the lumbosacral shear forces. Their estimation of 294±102 N of shear when carrying 30 kg in each hand may also suggest that current values may be high (McGill et al., 2013). However, the load carried in each hand would likely be held at or near the frontal plane containing the mediolateral axis of rotation, in contrast to the passenger and backpack loads which had greater moment arms. The closest shear value in the literature seems to be 483 N during a 22 kg stoop lift (Potvin et al., 1991), which was estimated for the L4-L5 joint and is an activity with a greater moment arm; the value is higher than walking with a 29 kg load which seems consistent.

The apparent risk caused by the shear lumbosacral joint reaction force when carrying 29 kg loads is minimal. The pars interarticularis, which supports much of the load, fails at higher loads of over 600 to 4000 N (Cyron, Hutton, & Troup, 1976; Lamy, Bazergui, Kraus, & Farfan, 1975). Much of the shear force is resisted by the intervertebral facet joints (Hall, 2007). The shear forces experienced when carrying a person or backpack may have stressed the supraspinous and interspinous ligaments as they strongly resist anterior shear during spinal flexion (Alexander, 1985). The deep erector spinae likely helped produce posterior shear forces for resisting the anterior shear forces due to sacral angle, load, and vertical ground reaction forces (Levangie & Norkin, 2005).

With the addition of a 284.5 N passenger or backpack, the shear joint reaction force increased by approximately 141 N; about 50% of the added load was expressed as a shear lumbosacral joint reaction force.

Lumbosacral Resultant Joint Reaction Force

The resultant lumbosacral joint reaction force exhibited the same pattern as the compressive joint reaction force; it increased from no load (287.08±30.50 N) to piggybacking (581.84±82.97 N) to backpacking (923.50±155.22 N). This increase could have been expected because the resultant lumbosacral joint reaction force must increase with the addition of a weight of a load, and an increase of 284.5 N of load would likely increase the joint reaction force more than any change of position would decrease it. Also, the change of position into greater trunk flexion created greater torques and therefore greater forces. This pattern would have also been expected mathematically, because the resultant joint reaction force is calculated using the compressive and shear joint reaction forces as its components in Pythagorean Theorem (Nordin & Frankel, 2012); the shear and compressive forces were both least for no load, the shear component was equal for the loaded conditions, and the compressive force increased from piggybacking to backpacking.

These results can be compared to results by Goh el al. (1998), who measured disproportionate increases of 26% and 64% in joint reaction force with added backpacks of 15 and 30 percent body weight. The loads ranged from 31 to 42 percent of body weight in the present study, which would be best compared with the 64% increase seen with the 30 % body weight load. However, substantially greater increases in resultant joint reaction force were seen here: 102% and 221% increases for piggybacking and backpacking respectively. The 29 kg load was 30% of body weight for two participants; their resultant joint reaction forces increased by 55% and 88% for piggybacking and 154% and 157% for backpacking. Khoo et al. (1995) found unloaded resultant lumbosacral joint reaction forces to be 1.73±0.28 times body weight, which can be compared to the 0.36±0.02 times body weight estimated here.

Note that Goh et al. (1998) found disproportionate increases in resultant JRF with increases in weight load; present results also showed disproportionate increases. This difference is likely due to the increased load coupled with the increased trunk flexion caused by load, which further increases torques and forces. The current values were greater than the 1998 study values, which may be due to model error within their model or the current model, or actual differences experienced due to the specific backpack and passenger used.

Some of the variation in the loads on the lumbar spine may have resulted from different lifting strategies, which can alter the proportion of load supported by each joint (Cholewicki et al., 1991). The internal lumbosacral joint reaction forces values are estimates provided by this biomechanical model and can be used as a guideline for further exploration.

Internal Forces and the Lumbar Spine

These results can be used to estimate the relative risk of each condition on the lumbar spine. Exhibiting the least amount of force for all three joint reaction forces variables, the no load condition placed the least amount of stress on the lumbar spine and may be considered the safest of these conditions. With theoretically equal amounts of shear force, the loaded conditions seem to be equal with respect to shear stress. The compressive force variable was the most informative with respect to risk to the lumbar spine. While carrying equal loads, carrying a person resulted in less than half of the compressive force of carrying a backpack; backpacking caused over double the amount of compressive force on the spine in comparison to piggybacking. When the shear and compressive joint reaction force values were combined to determine the resultant joint reaction force, piggybacking caused only approximately 60 % of the resultant lumbosacral joint reaction force caused by backpacking the same weight load.

Therefore, it seems reasonable to suggest that backpacking poses greater risk to the lumbar spine than does piggybacking. If training for a backpack trip, it would be advisable to begin with a piggybacking drill before proceeding to carrying a backpack of equal weight. A similar option would be to train with a front-back pack, which is more similar to piggybacking, to allow for a more natural gait and trunk position, or ideally, to use a front-back pack for the hike trip. If training for a competition that involves piggybacking, such as a CrossFit Games, it is recommended that athletes carry lighter passengers and progress to heavier passengers. If a backpack load is used for variety during training, a lighter backpack load should be used than would be used for piggybacking.

ALTERNATE METHOD AND VALUES FOR JOINT REACTION FORCES

The previous lumbar spine models that were followed only considered load anterior to the spine, producing forward torques. In the present model, the load was also posterior to the spine, adding backward torque. This created two problems. The traditional lumbar spine model sets the extensor muscle force negative in the y direction, adding to the compressive load. Therefore, the current model set the abdominal muscles in the opposite direction. However, although they may produce some tension at the posterior joint, the abdominal balloon theory is not supported and instead it is logical that they compress the joint overall. The second problem was accurately estimating the resulting muscle force.

The forward and backward torques may balance each other; as a result, this difference may not be the ideal representation of the muscle force occurring at the joint. The lumbar spine model would ideally be based on the total magnitude of muscle force, which was not measurable, and instead was based on a resultant that was able to make it appear that less torque was experienced when backpacking a 29 kg load (6 Nm) than when walking with no load (27 Nm). With this knowledge, the fact that the more accurately measured total magnitude of torque values were substantially greater for backpacking than the other conditions, because physics suggests that there should be more compressive force when carrying a load than when not carrying a load, a correction factor was created.

The muscle force values originally used in the equation were 460.98±107.08 N for no load, 362.80±181.80 N for piggybacking, and -99.40±219.39 N for backpacking based on 27.66±6.42 Nm, 21.77±10.91 Nm, and -5.96±13.15 Nm muscle torques. Instead of using these values and suggesting that the trunk flexors relieved the joint of 460 N when unloaded, for example, the muscle force values were equalized and all considered to compress the lumbosacral joint. To equalize the muscle force values, piggybacking was chosen as the unchanged values and the values for the other two conditions were multiplied by a correction factor. Backpacking muscle force values were multiplied by 3.65 (21.77/5.96) and no load values were multiplied by 0.78 (21.77/27.66). After multiplying by these conversion factors, it is suggested that the lumbar spine models for each condition started at comparable muscle values and that its values may be more accurate. The averages that would have been seen using this method are in Table 5.1.

Table 5.1: Hypothetical compressive, shear and resultant joint reaction forces, calculated using a muscle force correction factor and with the abdominal muscles adding to the joint reaction force.

Variable\Condition	No Load	Piggyback	Backpack	Bonferroni p-value NL/BP	NL/PB	BP/PB
Compressive L5-S1 JRF (N)	844.51±124.44	1092.00±208.99	1479.30±390.99	**.002**	**.000**	.083
Shear L5-S1 JRF (N)	272.24±27.43	413.59±27.43	413.59±27.43	**.000**	**.000**	1.000
Resultant L5-S1 JRF (N)	887.51±125.91	1168.90±203.21	1539.70±375.88	**.001**	**.000**	.085

- A Bonferroni post hoc test was used when the repeated measures ANOVA determined significance
- Statistically significant p-values are bolded, $\alpha = 0.05$

These compressive force values seem more realistic; the compressive forces are no longer extremely small for the no load condition and the compressive forces increase in the same order as the total magnitude of torque variable. However, the details of the lumbar spine model should be re-addressed and improved if used for future research. The external lumbosacral torque values may be a more appropriate method of assessing lumbar spine risk, unless it becomes possible to measure the total muscle forces experienced at the lumbosacral joint during these activities.

VERTICAL GROUND REACTION FORCE VARIABLES

The vertical ground reaction force patterns demonstrated the typical butterfly pattern with two major peaks and one interpeak dip (Robertson et al., 2004; Winter, 1991**)** during all three conditions. The very small variations or blips in the graphs were to be expected as every trial was different and a small change in movement, such as a change in arm swing or a hesitation as the participant tried to align with the force plate, could have caused an irregularity in the graph. These irregularities were evident on the graphs for individual trials, but were minimized when graphs were averaged.

A minor impact peak, as seen in Figure 2.22 of Chapter II, was anticipated just after footstrike as the foot first contacted the ground and absorbed impact. However, a defined impact peak was uncommon and irregular. Without a defined specific force value over a specific time frame, it was difficult to determine which small blips could be considered an impact peak. The overall impression was that impact peaks seemed to occur more often during unloaded walking, but a larger sample and a numerical definition of impact peak would be needed to determine significance.

When considering the relative vertical ground reaction force variables, it was evident that the gait pattern did not exhibit great variation between conditions. The first vGRF peak was slightly greater than the total weight of the system for all conditions, with means of 1.05 ± 0.06*TW for backpacking, 1.08 ± 0.09*TW for no load, and 1.09 ± 0.06*TW for piggybacking. These values were different between backpacking and piggybacking and the no load value was very close to the piggybacking value. This suggested that the participants were more comfortable taking each step and walked less cautiously with a passenger than with a backpack. This may reflect a more natural gait pattern during piggybacking, as the values were very close to no load values, and a more cushioned step during backpacking. This finding is supported by the results of the 2007 study by Birrell: moving load into a more centralized position on the trunk increased the magnitude of the vGRF impact peak.

The vGRF is known to stay close to body weight, fluctuating by approximately 30% of body weight (Robertson et al., 2004). The first peak was expected to be around 1.1 times body weight (Robertson et al., 2004). The relative vGRF interpeak minimum was approximately three quarters of the total weight. This is typical based on previous values, such as just under 0.8 to 0.85 (Roberston et al., 2004). The relatively small range of vertical forces between the first peak and interpeak dip suggests that the participants had minimal vertical accelerations, which corresponded visually to the video footage.

The second vGRF peak increased from the piggybacking (1.06±0.04 *TW) to the no load condition (1.12±0.05 *TW). Typical values are approximately 1.06 to 1.2 times body weight at the second peak (Robertson et al., 2004). This difference illustrated that although the gait pattern did not alter much by condition, the participants felt more comfortable to push off of the ground harder during the propulsive phase of the unloaded condition than for the piggyback condition. This is logical because an individual carrying a backpack or person may be concerned with the risk of dropping his bag or passenger, when an unloaded individual would only be concerned about himself and he would be accustomed to accelerating his own body. It may also be that rather than being concerned about their load, this difference was caused by gait alterations that made it easier for the participants to control the load.

The absolute vertical ground reaction force variables may provide more useful insight with respect to carrying these loads than the relative values because they displayed rather than masked the greater forces experienced when carrying a load. Force increases were seen from the unloaded to each loaded condition for all three vGRF variables, as expected (Birrell et al., 2007). With the addition of the 284.5 N person or backpack, the first vGRF peak increased by 320.88±81.65 N and 273.40±52.98 N respectively, the interpeak minimum increased by 181.31±37.83 N and 179.75±51.68 N respectively, and the second vGRF peak increased by 256.99±30.48 N and 281.65±45.92 N respectively. All of these increases were less than the magnitude of the weight of the added load, with the exception of the first peak during piggybacking; it seems reasonable that only a percentage of the added load is reflected within these vGRF values.

The two loaded conditions also demonstrated differences between absolute values for both the first and second vGRF peaks, both when the absolute vGRF values were compared with the ANOVA and Bonferonni tests and when the differences from the no load condition were compared with paired t-tests. It is interesting that although the added load was equal, the first peak had a greater increase from no load to piggybacking and the second peak had a greater increase from no load to backpacking. Based on the no load condition, the average participant had a stronger first peak during piggybacking than backpacking and had a more natural second peak during backpacking than piggybacking. Plots of the differences between vertical ground reaction forces from each load condition to the no load condition, as seen in Figures 4.29 and 4.31, further illustrated this point. With only a portion of the carried load reflected in the magnitude of the 1st vGRF peak for backpacking, the idea that increased knee flexion occurs for the purpose of "cushioning" the impact when carrying load was supported.

Equal loads suggested that approximately the same amount of vertical force would be experienced during the backpack condition if the vertical displacement was equal, but the participant may have exerted the impulse over a greater time period to allow for a lesser force exerted. The vertical displacement may not have been equal either and may be a variable of interest to examine in the future. The higher impact peak of piggybacking corresponds to the lesser angle of maximal knee flexion. During backpacking, the participants cushioned their impact, demonstrated by a greater knee flexion angle and a lesser first peak of the vertical ground reaction force. Also, the second peak is experienced when the body mass is primarily being moved forward, but also upwards, so the load would not necessarily produce a proportional increase in vGRF**.**

Comparing the first peak results between the relative and absolute values addresses the causes of these differences. Because the relative first peaks were only different between loaded conditions but the absolute first peaks were different between all conditions, it is evident that the differences between no load and each load condition were due to the added load. Because both the relative and absolute first peaks were different between the two loaded conditions, which involved carrying equal weights, the differences between the piggybacking and backpacking first vGRF peaks must have been because of changes in gait pattern.

The piggyback condition demonstrated a mean of 113% of the load weight as vGRF during the first peak and only 90% during the second peak. The backpack condition demonstrated values of 96 % and 99% respectively, which were both closer to 100%. With further research, it may be seen that backpacking vertical ground reaction forces are closer to the natural gait pattern.

There was a proportional increase in vGRF to added load. This was also noted by Birrell et al. (2007) and Lloyd et al. (2011).

GAIT VARIABLES

Contrary to expectations, the only two gait variables that demonstrated differences between conditions were right foot stance time and double stance time. Walking velocity and cadence did not increase with load, average step length did not decrease with load, right to right and left to left stride lengths did not decrease with load, and single stance time did not increase with load. These variables may have shown significant differences if their measurements were to have been obtained with greater accuracy. Timing gates would be recommended rather than on-screen methods for future measurements of walking velocity. Step and stride lengths would have been better measured by a pressure-sensing mat. Stance time was best measured by the force plate, which operated at 200 Hz, rather than by video at 30 Hz. This is evident when comparing the single stance time variable measured by video frame, which was not found to be significant, to the right foot stance time measured by the force plate data.

The right foot stance time as measured by the force plate was different between all three conditions. The duration that the foot was in contact with the force plate increased from 0.714±0.040 s with no load to 0.747±0.034 s with a passenger, to 0.766±0.037 with a backpack. Other researchers (Birrell et al., 2007) have noted increases in stance time when carrying load, so this result was expected. One explanation for this phenomenon is that individuals increase the time that the feet are in contact with the ground to increase stability. When load is added posterior to the spine, the centre of gravity shifts posteriorly; the greater the shift, the greater the decrease in stability (Birrell et al., 2007; Chow et al., 2005). This may also have been because of the added danger of falling with a load, caution used to prevent injury to the load, or because the individual's sense of balance and proprioception was truly challenged by the added load. The piggyback right foot stance time was closer to the no load stance time, suggesting that less ground contact time was needed because the participants felt more stable when carrying a person than when carrying a backpack. This finding is supported by Birrell et al. (2007); stance duration increased with load, but not by as much as when the load was transferred to a more centralized position on the trunk. Gillet et al. (2006) also noted that a more natural gait was demonstrated when a proportion of a posterior load is moved to the anterior trunk. These findings, in conjunction with previous literature, support the theory that piggybacking gait is closer to natural gait than is backpacking gait.

If carrying load above the head is considered similar to piggybacking because the centre of gravity of the load is better aligned with the centre of gravity of the carrier, then these findings are also supported by the 2011 results of Lloyd et al. Stance time increased from carrying no load to carrying the load superior to the head to carrying a load posterior to the spine (Lloyd et al., 2011).

Right foot stance time was seen to increase when measured at 200 Hz so it is assumed that left foot stance time would have also increased, and these increases would have led to an increase in double stance time. If this assumption is true despite the lack of significant data, the 30 fps video data likely was not sufficiently frequent to identify these changes. However, double stance time did increase from no load to backpacking, which further illustrates the need for longer ground contact time while carrying a backpack (Birrell et al., 2006; Chow et al., 2005; Cottalorda et al., 2003).

One substantial difference between backpacking and piggybacking that may affect both right foot stance time and double stance time is the activity of the load. The piggyback load is a person who is actively holding on and has some control over their movement. If there happens to be a lot of transverse plane motion, for example, the passenger may pull closer to the trunk of the carrier to decrease the torques about the longitudinal axis, while the backpack would maintain the same strapped on position. Also, as it was shown earlier in the kinematic and torque variables sections, backpacking produces greater biomechanical alterations from unloaded gait than does piggybacking. These changes may also contribute to the increase in right foot stance time from piggybacking to backpacking and the difference in double stance time from no load to backpacking but not to piggybacking.

QUALITATIVE OBSERVATIONS AND COMMENTS

The participants and researchers were able to form qualitative opinions comparing the loaded conditions based on participant comments, pilot study results, and personal experiences during this study. Several participants made comments that they thought the backpack was heavier than the passenger; they were shocked to learn that the loads were of equal weight. During the pilot project, one carrier explained that he much preferred piggybacking a 50 kg woman to carrying a 25 kg backpack. Also, the researcher helped the participants lift and put on the backpack because it was noticeably heavy, whereas physically helping mount the passenger seemed unnecessary. These findings are intuitive because many people would carry a person of the same mass of themselves or possibly even heavier, which may be 100 kg, and many people would not consider it dangerous. Therefore, a 29 kg child is comparably a very light weight load. Considering that the 29 kg backpack was noticeably heavy and mounted with caution, it is highly unlikely that the same individual who can carry a 100 kg person could carry a 100 kg backpack. This reasoning leads one to conclude that a backpack is more strenuous to carry than a person of equal weight, and the various biomechanical differences in load explored during this study help to understand the mechanisms of this difference.

APPLICATIONS

According to the variables and values explored in the present study, both load conditions demonstrated several changes from unloaded walking and backpacking represents a greater biomechanical deviation from natural gait. Because such deviations included a greater trunk flexion angle while loaded, greater total magnitudes of torque, greater muscle forces, and increased vertical ground reaction forces and because of the extent of these deviations, the injury risk would be lowest for the no load condition and highest for the backpack condition. As noted, pain or injury to the back is very common; improper load carriage is likely a contributor. If people are aware that there may be risk involved in carrying a loaded backpack or even a person, the hope is that they would provide some thought to the technique and loads which they are lifting or carrying.

Because there are greater biomechanical alterations during backpacking, it seems that the risk to the lumbar spine can be minimized by carrying the same weight of person instead, or by moving some of the backpack load anterior to the trunk. This way, an equal load can be used to train the strength of the lower extremity muscles while exposing the lumbar spine to less risk. If training to carry a person, it is recommended that a safe progression is followed and other than for variety or the convenience of creating the load weight, there would be no obvious benefit to training with a backpack load first. However, if training to carry a heavy backpack, it would be beneficial to alternate training with a loaded backpack and with a person. This would allow the same amount of load to be experienced and adapted to by the lower body while decreasing the stress on and altered position of the trunk.

CHAPTER VI

SUMMARY, CONCLUSIONS, AND RECOMMENDATIONS

SUMMARY

Piggybacking is commonly used as a strength training drill in CrossFit, bootcamps, team sport training (Associated Newspapers, 2011; Donohue, 1996; Nottingham Rugby, 2011; Payson Roundup, 2013), fitness competitions and races (Madliger, 2012; Muller, 2013), and partner exercise plans found in popular magazines (Schneider, 2009). Most people have performed a piggyback carry (Bennett, 1999; Lileks, 2007; YMCA, 2012), yet the biomechanics of the exercise and considerations relevant to risk to the spine have often been overlooked. With only backpacking literature available to provide a background for expectations when carrying a posterior load, the handful of front-back pack studies (Gillet et al., 2006; Kinoshita, 1985) were most relevant.

The purpose of this study was to investigate the kinematic differences in maximal trunk flexion angle during gait while unloaded, piggybacking and backpacking. It was hypothesized that the maximal trunk flexion angle for piggybacking would be greater than for no load and different than for backpacking. Secondary purposes included the investigation of differences in kinematic, resultant load torque, internal lumbosacral joint torque and force, vertical ground reaction force, and gait variables. It was hypothesized that the values for these variables would not be equal across conditions. During the analysis, additional external lumbosacral torque variables were added to better explain the differences in trunk flexion angle and the general biomechanics of each condition.

Twelve strength-trained adult males with an average mass of 82.4±8.3kg participated in the study, which was conducted at the Pan Am Clinic Foundation Laboratory on April 8, 2014.

Each participant walked over an embedded force plate while being video-recorded with three cameras for three successful trials per condition: unloaded, piggybacking, and backpacking. Video data was analyzed using 2014 Dartfish Software 6 TeamPro 6.0 and the Advanced Mechanical Technology, Inc (AMTI) force plate data was analyzed using Microsoft Excel 2010.

Means and standard deviations were compared among the three conditions using Repeated-Measures Analysis of Variance tests with a Bonferroni correction when differences were identified. A significance level of $p<0.05$ was used to identify significant differences in these means. Kinematic variables were measured at the positions of maximal angles and lumbosacral load variables were measured at the position of maximal trunk flexion. The vertical ground reaction force was measured over the course of a footstep; the peaks and minimum values were identified and averaged for each participant, and compared between participants. All gait variables were measured for distance and/or length of time using the video data, except for right foot stance time which was also measured using the force plate data.

Biomechanics of Piggybacking

The results of this study suggested that a posterior load did in fact result in compensatory trunk flexion (Goh et al., 1998). The amount of compensatory trunk flexion seemed to depend on weight of the load and the location of the centre of gravity of the load; the heavier the load and the more posterior the load, the greater trunk flexion to be expected. Although the piggyback and backpack loads were of equal weight, piggybacking resulted in less trunk flexion due to a more anterior centre of gravity of the load. Trunk flexion angle may be the most important characteristic to measure when considering risk to the lumbar spine or gait alterations. The magnitude of trunk flexion seemed to be a result of the balancing of lumbosacral torques that are important determinants of the stress placed on the lumbosacral joint, and it is therefore an obvious and meaningful representation of the stress due to a specific load.

Piggybacking Gait

As demonstrated by several variables, piggybacking gait was biomechanically more similar to unloaded gait than was backpacking gait. Some of these variables were maximal trunk flexion angle, trunk flexion-extension range of motion, maximal knee flexion angle, torque caused by the body of the carrier, total resultant torque, total magnitude of torque, the percentages of negative versus positive torque, the compressive and resultant lumbosacral joint reaction forces, the first relative vertical ground reaction force peak, and right foot stance time. However, the mean piggybacking second relative vertical ground reaction force value was not between the no load and backpacking values.

CONCLUSIONS

The following conclusions have been determined based on the results of this study:

- Trunk flexion angle increased from walking with no load to piggybacking 29 kg to backpacking 29 kg.
- While carrying load, the trunk remained somewhat flexed throughout the gait cycle. The trunk range of motion was similar for no load and piggybacking, but almost doubled for backpacking.
- The first vertical ground reaction force peak was greater for piggybacking than backpacking, which may correspond to the greater knee flexion during backpacking.
- The piggyback load produced greater resultant torque and total magnitude of torque than the backpack load.

- The body of the carrier produced a forward-rotating torque for all conditions, which increased in magnitude from no load to piggybacking to backpacking.
- The total resultant torque, produced by the load and the body of the carrier, was backward-rotating for backpacking and forward-rotating for piggybacking and no load. Therefore, the erector spinae dominated during no load and piggybacking and the rectus abdominus dominated during backpacking.
- The total magnitude of torque, produced by the load and the body of the carrier, increased from no load to piggybacking to backpacking.
- Backpacking most effectively balanced its forward- and backward-rotating torques.
- The partly anterior distribution of load during piggybacking allows a more natural gait pattern than backpacking.

RECOMMENDATIONS

Future studies should:

- Continue to improve the lumbar spine model, by considering using a different moment arm length when the trunk flexors dominate and by trying to account for the absolute torques rather than basing the joint reaction forces on a resultant value. Electromyography may provide a better idea as to the proportions of trunk flexor versus extensor forces.
- Consider using an inverse dynamics model, to account for the larger range of motion and accelerations during backpacking as compared to carrying no load and piggybacking.

- Utilize a three dimensional motion capture system if available, in order to account for any motion in the third dimension and decrease measurement error.
- Measure the range of motion in the transverse plane and investigate if transverse plane motion decreases while sagittal plane motion increases with a piggyback load.
- Investigate the biomechanics of females carrying passengers. Through the current research, it has become apparent that piggybacking is more commonly used by females. When given the choice in CrossFit competition, males often choose the fireman's carry.
- Compare the biomechanics of the different methods of carrying a person.
- Include a variety of passengers for piggybacking in order to determine how much the strength and effort of the passenger effects that position of the trunk and resulting forces and torques.
- Include a variety of different weight loads to increase the generalizability of the results and to comment on the effect of load on any differences between piggybacking and backpacking.

Questions to Continue Exploring:

- Should the piggyback be performed, by whom and while carrying whom? What relative weights should be considered acceptable and is this too strenuous for a subset of the population?
- What progressions should be advised and recommended against?

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APPENDIX A

Ethics Approval

UNIVERSITY OF MANITOBA | Research Ethics and Compliance
Office of the Vice-President (Research and International)

Human Ethics
208-194 Dafoe Road
Winnipeg, MB
Canada R3T 2N2
Phone +204-474-7122
Fax +204-269-7173

APPROVAL CERTIFICATE

March 17, 2014

TO: **Sheena Graham** (Advisor M. Alexander)

FROM: [redacted]

Re: **Protocol #E2014:021**
"Biomechanical Comparisons Considering Risk to the Lumbar Spine: Walking with No Load, a Backpack, and a Person on the Back"

Please be advised that your above-referenced protocol has received human ethics approval by the **Education/Nursing Research Ethics Board**, which is organized and operates according to the Tri-Council Policy Statement (2). **This approval is valid for one year only**.

Any significant changes of the protocol and/or informed consent form should be reported to the Human Ethics Secretariat in advance of implementation of such changes.

Please note:

- **If you have funds pending human ethics approval, please mail/e-mail/fax (261-0325) a copy of this Approval (identifying the related UM Project Number) to the Research Grants Officer in ORS in order to initiate fund setup. (How to find your UM Project Number: http://umanitoba.ca/research/ors/mrt-faq.html#pr0)**

- **If you have received multi-year funding for this research, responsibility lies with you to apply for and obtain Renewal Approval at the expiry of the initial one-year approval; otherwise the account will be locked.**

The Research Quality Management Office may request to review research documentation from this project to demonstrate compliance with this approved protocol and the University of Manitoba *Ethics of Research Involving Humans*.

The Research Ethics Board requests a final report for your study (available at: http://umanitoba.ca/research/orec/ethics/human_ethics_REB_forms_guidelines.html) **in order to be in compliance with Tri-Council Guidelines.**

PANEL ON
RESEARCH ETHICS
Navigating the ethics of human research

TCPS 2: CORE

Certificate of Completion

This document certifies that

Sheena Graham

has completed the Tri-Council Policy Statement:
Ethical Conduct for Research Involving Humans
Course on Research Ethics (TCPS 2: CORE)

Date of Issue: **8 February, 2014**

UNIVERSITY
OF MANITOBA

The University Of Manitoba

Certifies that

Sheena Graham

Completed the

Research Integrity Online Course: Biomedical Sciences

January 29, 2014

APPENDIX B

Consent Forms

Informed Consent Form - Participant

Research Project Title: Biomechanical Comparisons Considering Risk to the Lumbar Spine: Walking with No Load, a Backpack, and a Person on the Back

Principal Investigator and Contact Information:
Sheena Graham, umgraha3@cc.umanitoba.ca

Research Supervisor and Contact Information:
Dr. M. Alexander, alexan@cc.umanitoba.ca, 474-8642

This consent form, a copy of which will be left with you for your records and reference, is only part of the process of informed consent. It should give you the basic idea of what the research is about and what your participation will involve. If you would like more details about something mentioned here, or information not included here, you should feel free to ask. Please take the time to read this carefully and to understand any accompanying information.

1. The purpose of this research is to determine if there are biomechanical differences in the postural alignment and/or lumbosacral forces experienced by male CrossFit athletes while walking when a human load or backpack load is added. The human load will be a child of approximately 29 kg that participants will carry using the piggyback. This may help identify which type of load places more stress on the lower back.

2. The participant will be asked to briefly note their experience performing the piggyback carry, carrying a heavy backpack, and carrying a heavy load of any other nature. The participant will also be asked to disclose any previous history of musculoskeletal injury, because an injury may be cause for the individual to withdraw from this study. They will then be asked to sign this consent form if choosing to participate in the study. The participant will choose an envelope at random, which will determine the order of the loaded conditions. First, height and weight will be measured on a scale and then markers will be taped to 8 locations on the body. Specific landmarks will be palpated on the body and several body segments will be measured in order to locate the appropriate position for each marker. The participant will have several minutes to practice walking so that the full right foot, and only the right foot, contacts the force plate while moving along the walkway. While data is recorded, each participant will be asked to walk down the 8 meter walkway until three successful trials are recorded for each of the three conditions. During each trial, the video cameras and force plate will be activated and the participant will walk at a natural pace from start to finish. The participant will have a loaded backpack lifted onto the back and fastened around the chest and pelvis for the backpack condition. When both the child and participant agree that they are ready, the participant will crouch moderately while the child hops onto their back for the piggyback condition. Please be aware of the safety and welfare of the piggybacked child throughout testing. Please let the researchers know of any concern you have for the child immediately if you should feel concern.

The participant will be able to perform a couple of practice trials for each condition. It is expected that each participant will only be needed for one visit, lasting approximately 1 to 2 hours.

3. Three video cameras, two Canon D2L cameras and one Canon HDV 1080i camera, will record video during the trials, from one left and two right side views. The right foot will land on the force plate for one foot step per trial. This will measure the ground reaction forces for the duration that the foot is in contact with the ground.

4. Volunteering for this study will provide a unique experience to the participants, with a glimpse into current research at the biomechanics laboratory at the Pan Am Clinic Foundation. There is also the chance that any unusual gait or movement biomechanics will be noted during data analysis, which may lead to a decrease in future risk of injury. The participants will help the researchers to address the existence and/or magnitude of low back stress experienced during these loaded conditions. Once the study has concluded, participants will be provided with an explanation of results that can be applied to carrying a load during CrossFit competition, training programs or everyday life.

5. Because the participants already perform the piggyback carry or another skill with much heavier loads during CrossFit practice, and because the position of load is somewhat similar to within a loaded backpack, there should be no additional risk during this study.

6. The data collected will be confidential. Each participant will be assigned a number for identification. It is not possible for the data to be completely anonymous because the videos will identify the participants and they need to be matched with the force plate data for analysis purposes. The videos and force plate data will be stored on a university computer with a password in a private lab. Only the principal investigator will have access to this data, and may access the data anytime from the date of data collection until January 2015. The research advisor may also view or access the data upon request. The principal investigator will hopefully stop needing access to the data by August 2014, but would like to be able to have access until the thesis work has been finalized and passed. If photos are extracted from the videos to be included in the thesis, the participant's face will be blocked for anonymity. The data will be deleted by January 2015.

7. Participation in this study will be voluntary.

8. The participant may withdraw from this study at any point in time, and should notify the principal investigator as soon as possible if he should choose to withdraw. Any data collected on such a participant will be deleted.

9. The biomechanics of load carriage will be analyzed with respect to the purpose of the study, as seen in 1. After all of the data has been collected, a written debriefing explanation will be emailed to the participants. In approximately January 2015, a summary of results will also be emailed to the participants.

10. The results of this study will be included in a MSc Thesis, which will be included on the University of Manitoba's Thesis electronic database. Copies will also be printed for each thesis committee member and the University of Manitoba's Biomechanics Laboratory. If there is the opportunity, the results of this study may also be published in scientific journals, presented at relevant conferences, or disseminated by other means.

11. A brief (1-3 pages) summary of the study and results will be emailed to participants in approximately January 2015.

12. All confidential data will be destroyed by January 2015.

Your signature on this form indicates that you have understood to your satisfaction the information regarding participation in the research project and agree to participate as a subject. In no way does this waive your legal rights nor release the researchers, sponsors, or involved institutions from their legal and professional responsibilities. You are free to withdraw from the study at any time, and /or refrain from answering any questions you prefer to omit, without prejudice or consequence. Your continued participation should be as informed as your initial consent, so you should feel free to ask for clarification or new information throughout your participation.

The University of Manitoba may look at your research records to see that the research is being done in a safe and proper way.

This research has been approved by the Education and Nursing Research Ethics Board. If you have any concerns or complaints about this project you may contact any of the above-named persons or the Human Ethics Coordinator (HEC) at 474-7122 (Margaret.Bowman@umanitoba.ca). A copy of this consent form has been given to you to keep for your records and reference.

--- Signatures as Required-------------------------------------

Participant's Signature ____ Date ____________

Researcher and/or Delegate's Signature ____________________________ Date ____________

Informed Consent Form – Parent or Guardian of Child

Research Project Title: Biomechanical Comparisons Considering Risk to the Lumbar Spine: Walking with No Load, a Backpack, and a Person on the Back

Principal Investigator and Contact Information:
Sheena Graham, umgraha3@cc.umanitoba.ca

Research Supervisor and Contact Information:
Dr. M. Alexander, alexan@cc.umanitoba.ca 474-8642

This consent form, a copy of which will be left with you for your records and reference, is only part of the process of informed consent. It should give you the basic idea of what the research is about and what your participation will involve. If you would like more details about something mentioned here, or information not included here, you should feel free to ask. Please take the time to read this carefully and to understand any accompanying information.

1. The purpose of this research is to determine if there are biomechanical differences in the postural alignment and/or lumbosacral forces experienced by male CrossFit athletes while walking when a human load or backpack load is added. Your child will be asked to be the piggybacked human load of approximately 29 kg. This study may help identify which type of load places more stress on the lower back.

2. When you arrive, you will be asked to fill out this form and your child will be asked to read and sign the Assent form. Your child's height and weight will be measured on a scale. Markers will be taped to your child at 11 locations. Specific landmarks will be palpated on the body and several body segments will be measured in order to locate the appropriate position for each marker. The marker positions are included in Figure 1 below, plus additional markers on the centre of gravity of the thigh, lower leg, and foot. During data collection, each participant will be asked to walk down the 8 meter walkway until three successful trials are recorded for each of the three conditions. During each trial, the video cameras and force plate will be activated and the participant will walk at a natural pace from start to finish. The participant will have a loaded backpack lifted onto the back and fastened around the chest and pelvis for the backpack condition. For the piggyback condition, the participant will crouch moderately and your child will be asked to hop carefully onto the back of the participant. Your child will be asked to hold onto the participant around the shoulders and neck without putting pressure on the neck. Please remind your child to let you or the researchers know immediately if he/she is uncomfortable or slipping down the back of the participant. The participant will be able to perform a couple of practice trials for each condition. It is expected that your child will only be needed for one visit, but may be asked to participate in a second or third visit if there is a shortage of participants that are available on the day of testing. You, or another adult that you choose to have supervise your child, and your child are expected to be needed for the duration of the data collection. This will likely be for several hours, on at least one date.

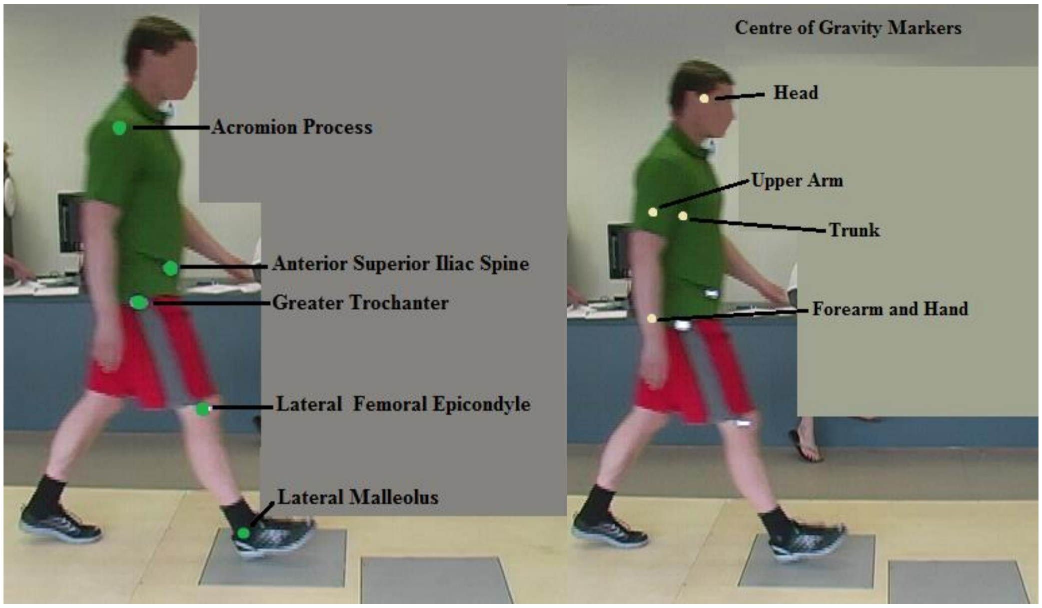

Figure 1: A) Green tape will identify the bony landmarks and B) Beige tape will identify the location of the centre of gravity of important limb segments.

3. Three video cameras, two Canon D2L cameras and one Canon HDV 1080i camera, will record video during the trials, from one left and two right side views. The participant's right foot will land on the force plate for one foot step per trial. The force plate will measure the ground reaction forces, which are a result of the weight of the participant and the child together, for the duration that the foot is in contact with the ground.

4. Volunteering for this study will provide a unique experience to you and your child, with a glimpse into current research at the biomechanics laboratory at the Pan Am Clinic Foundation. Your participation will help the researchers to address the existence and/or magnitude of low back stress experienced during these loaded conditions. Once the study has concluded, participants and the public will be provided with an explanation of results that can be applied to carrying a load during CrossFit competition, training programs or everyday life.

5. Because there is minimal risk to the passenger in a piggyback carry, there should be minimal risk to your child during this study. Often, children are already passengers in piggybacking in everyday life. Also, the child will be encouraged to vocalize any issues or comfort or slippage throughout the study and can withdraw at any time.

6. The data collected will be confidential. Each participant will be assigned a number for identification. The child will be referred to as the PB Load (piggyback load). It is not possible for the data to be completely anonymous because the videos will identify the participants and they need to be matched with the force plate data for analysis purposes. The videos and force plate data will be stored on a university computer with a password in a private lab. Only the principal investigator will have access to this data, and may access the data anytime from the date of data collection until January 2015. The research advisor may also view or access the data upon request. The principal investigator will hopefully stop needing access to the data by

August 2014, but would like to be able to have access until the thesis work has been finalized and passed. If photos are extracted from the videos to be included in the thesis, the participant's face will be blocked for anonymity. The data will be deleted by January 2015.

7. Participation in this study will be voluntary.
8. Your child may withdraw from this study at any point in time, and should notify the principal investigator as soon as possible if he should choose to withdraw. Any data collected using your child will then be deleted if requested.

9. The biomechanics of load carriage will be analyzed. After all of the data has been collected, a written debriefing explanation will be emailed to you. In approximately January 2015, a summary of results will also be emailed to you.

10. The results of this study will be included in a MSc Thesis, which will be included on the University of Manitoba's Thesis electronic database. Copies will also be printed for each committee member and the University of Manitoba's Biomechanics Laboratory. If there is the opportunity, the results of this study may also be published in scientific journals, presented at relevant conferences, or disseminated by other means.

11. A brief (1-3 pages) summary of the study and results will be emailed to you in approximately January 2015.
12. All confidential data will be destroyed by January 2015.

Your signature on this form indicates that you have understood to your satisfaction the information regarding participation in the research project and agree to allow your child to participate. In no way does this waive your legal rights nor release the researchers, sponsors, or involved institutions from their legal and professional responsibilities. Your child is free to withdraw from the study at any time, and /or refrain from answering any questions that you prefer to omit, without prejudice or consequence. Your continued participation should be as informed as your initial consent, so you should feel free to ask for clarification or new information throughout your participation.

The University of Manitoba may look at your research records to see that the research is being done in a safe and proper way. This research has been approved by the Education and Nursing Research Ethics Board. If you have any concerns or complaints about this project you may contact any of the above-named persons or the Human Ethics Coordinator (HEC) at 474-7122 (Margaret.Bowman@umanitoba.ca). A copy of this consent form has been given to you to keep for your records and reference.

---Signatures as Required---------------------------------

Parent or Guardian's Signature ___________________________________ Date ____________
(of the child to be piggybacked)

Researcher and/or Delegate's Signature ____________________________ Date ____________

Assent Form - Child

Research Project Title: Biomechanical Comparisons Considering Risk to the Lumbar Spine: Walking with No Load, a Backpack, and a Person on the Back

Principal Investigator and Contact Information:
Sheena Graham, umgraha3@cc.umanitoba.ca

Research Supervisor and Contact Information:
Dr. M. Alexander, alexan@cc.umanitoba.ca 474-8642

A copy of this form will be left with you to give to your parent or guardian. It is important for you to read this form in order to understand what this study is asking you to do. If you have any questions, please ask the researcher or your parent or guardian.

1. The purpose of this research is to see if there are biomechanical differences in the forces on the back or the way a person walks between piggybacking a person and carrying a backpack. You have been asked to be the person getting a piggyback ride because this study needed someone that weighed as much as you. This study may help figure out whether it is more risky for the carrier to carry you, or a backpack that weighs the same as you.

2. When you arrive, you will be asked to read and sign this form. You will stand on a scale and the researcher will measure your height and weight. Markers will be taped to you at 11 spots. The researcher may need to find certain bones on your body, using their hands, in order to know where to place the markers. The researcher will be careful when placing the markers but you can tell them to stop if you feel uncomfortable. The pictures below show where the tape markers will be placed, and there will also be markers on your thigh, lower leg, and foot. During the experiment, each participant will need to walk the runway with you on their back at least three times. They might also want to do a couple of practice walks first. During each trial, the video cameras and force plate will be activated and the participant will walk at a natural pace from start to finish.

When you are about to piggyback, the participant will crouch down and you will be asked to hop carefully onto the back of the participant. You will be asked to hold onto the participant around the shoulders and neck without putting pressure on the neck, because you do not want to choke the participant. Try to hold onto every participant in the same way if possible. Please tell your parent, guardian, or the researchers if you are uncomfortable or slipping down the back of the participant. You might only need to come help with this study on one day, but may be asked to come back for a second or third visit if there were not enough participants on the first day of testing. You and the adult who is supervising you, which will likely be your parent or guardian, will be needed for the whole experiment. This will likely be for several hours, on at least one date.

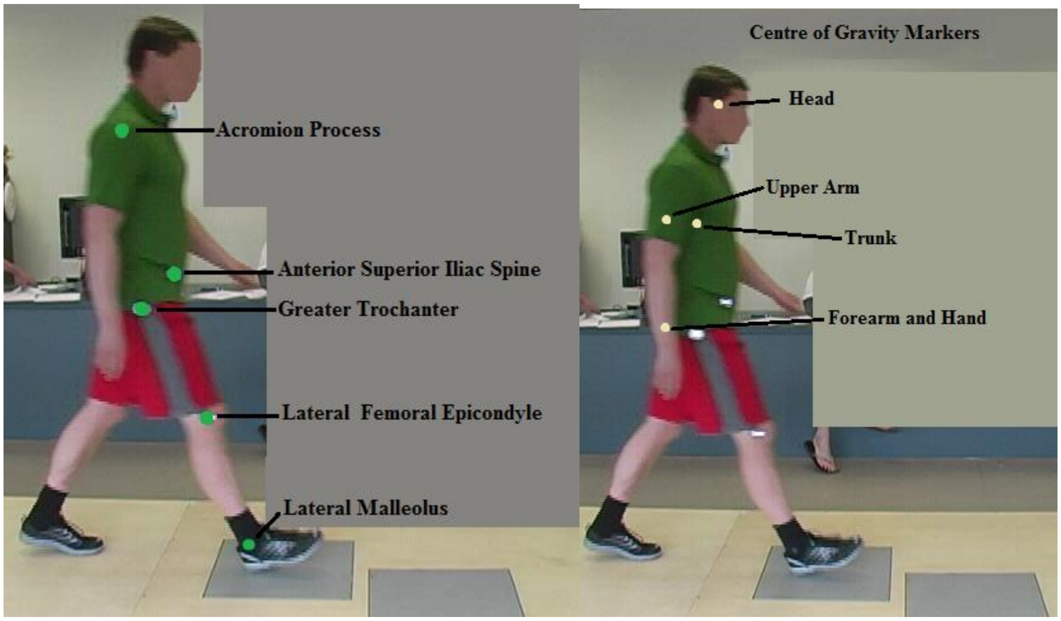

Figure 1: A) Green tape will identify the bony landmarks and B) Beige tape will identify the location of the centre of gravity of important limb segments.

3. Three video cameras will record video during the experiment. One will be on your left side and two will be on your right side. The participant's right foot will land on the grey square on the floor, called the force plate, as they walk down the walkway. The force plate will measure the forces from the ground when it is stepped on.

4. Volunteering for this study will show you what it is like to be part of research and what a biomechanics laboratory is like. Your participation will help the researchers to see if backpacks or piggybacks might be harder on the back of the carrier. Once the study has finished, the public will get an explanation of the results that might help teach people to piggyback more safely in competition, training programs or everyday life.

5. There is a very small amount of risk to you during the piggyback carry. You will need to make sure that you tell the carrier, the researcher, or your parent or guardian that you are slipping down if you feel like you are falling down the back of the carrier, or tell them if you are uncomfortable. Before you jump on the back of the carrier, you need to wait for the carrier to tell you that he is ready. Once he tells you he is ready, he will catch you and you will hold onto him around his shoulders. If you need to, you may also stand on a step in order to load onto the back of any carrier. You have probably been piggybacked in everyday life and know what is going on, but the researchers want to keep you safe so be sure to tell somebody if you feel like you are slipping or if you need to stop.

6. All of the videos and information collected will be confidential. You will not be referred to by name, but it is not possible for the data to be completely anonymous because the videos will show your faces. The videos and force plate data will be stored on a university computer with a password in a private lab. Only the main researcher will be able to see this information, and may access the data anytime from the date of the experiment until January 2015. She will hopefully

stop needing to see the data by August 2014, but would like to be able to have access until the thesis work has been passed. If photos are taken from the videos to be included in the thesis, your face will be blocked just like in the picture above.

7. Participation in this study will be voluntary.
8. You may stop this study at any point in time, and should tell the main researcher as soon as possible if you need to quit. Any data collected using you will then be deleted if you want it to be.

9. The researchers will use the videos and force plate information to see if there are differences between carrying you and carrying a backpack. After at least 12 participants have piggybacked you, your parent or guardian will get an email explaining the study. By January 2015, a second and longer email will explain the results of the study.

10. The results of this study will be included in a MSc Thesis, which is like a book on the study. It will be included on the University of Manitoba's Thesis electronic database. Copies will also be printed for each study supervisor and the University of Manitoba's Biomechanics Laboratory. The results of this study may also be published in scientific journals, presented at relevant conferences, or shown to people in other ways.

11. A brief (1-3 pages) summary of the study and results will be emailed to you in approximately January 2015.
12. All confidential data will be destroyed by January 2015.

Please sign this form if you understand the information about what you will be asked to do during this study and if you would like to be the passenger of several piggyback rides. In no way does this waive your legal rights nor release the researchers, sponsors, or involved institutions from their legal and professional responsibilities. If you would like, you can stop being part of this study at any time, without prejudice or consequence. If you ever are unsure about what is going on, feel free to ask for clarification or new information throughout the experiment.

The University of Manitoba may look at your research records to see that the research is being done in a safe and proper way. This research has been approved by the Education and Nursing Research Ethics Board. If you have any concerns or complaints about this project you may contact any of the above-named persons or the Human Ethics Coordinator (HEC) at 474-7122 (Margaret.Bowman@umanitoba.ca). A copy of this consent form has been given to you to keep for your records and reference.

---Signatures as Required---------------------------------

Child (to be piggybacked)'s signature ______________________________Date____________

Parent or Guardian's Signature ___________________________________ Date ____________
(of the child to be piggybacked)

Researcher and/or Delegate's Signature ____________________________ Date ____________

APPENDIX C

Other Forms

EMAIL BULLETIN

Hello CrossFit!

My name is Sheena Graham and I am a Graduate Student at the University of Manitoba. For my thesis project, I am doing a Biomechanical analysis of load carrying. I will be comparing the kinetics and kinematics of carrying a backpack with the piggyback (example photo: http://www.flickr.com/photos/thor604/7672449646/).

I have found many pictures of CrossFit athletes using the piggyback to train, on various internet sites (example: http://crossfitmotivate.com/2013/05/13/may-13-2013/), and am hoping to:

*A) Recruit willing CrossFit athletes who currently train using the piggyback as participants for my research study. I understand that there may be a lack of qualifying individuals in Winnipeg, and may need to accept CrossFit athletes who train while carrying any type of load on the back, or a heavy load in general.

> I will be creating a list of interested candidates by March 20, 2014, which will include the names and contact information of athletes that email me expressing interest. I will ask for the participants' time in either late March or April 2014. All data should be able to be collected in one visit, which might range from 1 hour to 2 hours. This will occur at the Pan Am Clinic Biomechanics Lab on a day that the lab is available to us.

B) Gain an idea as to who uses the piggyback carry for strength training in Winnipeg.
I would appreciate any feedback you have for me. If possible, I would ask that you email me back with any information you have on the following (umgraha3@cc.umanitoba.ca):

> 1) Who trains using the piggyback at your facility? (if not, do you carry another person on the back in a different position or do you ever use a backpack?)
> 2) What you use as a weight load when carrying another person – a person of the same weight, or someone who weighs half as much as the carrier, for example, or what is the criteria?

* I am attaching a bulletin in hopes that you may spread the word to your members and help me find interested participants. Feel free to email it to your members, post it on your website or blog, or pass it through the Winnipeg CrossFit community in order to help this bulletin reach various interested CrossFit athletes.

Thank you very much for your time! I look forward to hearing from you,

Sheena Graham
BSc KIN (Exercise Science)
CSEP-CPT

Have YOU ever been biomechanically analyzed?

Would you like to participate in a Piggybacking and Backpacking Study?!

If you: Are a male CrossFit athlete over 70 kg
Regularly train with heavy loads
Possibly use the Piggyback (not required)

You may be included in a research study in which your walking and carrying will be biomechanically analyzed!

Only 1 session needed!

Benefits:

- Your data and results will be available to you once the study has concluded
- Unusual gait or movement biomechanics may be identified
- Gain a glimpse into the research performed at the Pan Am Clinic Foundation Biomechanics Lab
- Volunteer experience, especially valuable if pursuing work or education in the field of Kinesiology or Medicine

If interested, please contact umgraha3@cc.umanitoba.ca

A late February testing date is expected, please apply ASAP

INTRODUCTORY EMAIL - Participant

Hello ______,

Thank you for expressing interest in this study of load carriage. You have been added to the list of participants that may be included in this study, in the order of when they first contacted me. Once the date is set for testing, I will contact the first 12 participants to see if they are available at the requested time. If any participants are not able to attend, the individuals next on the list will be contacted.

It is expected that you will only be required to come to the Pan Am Clinic Foundation Biomechanics Laboratory on one day, likely for approximately 1 to 2 hours. This will likely be in late March or early April. I have also attached 1) the Instructional Email and 2) the Informed Consent Form, which provide further information on what you are to expect if you choose to participate.

To ensure that you are eligible to participate, please read through the inclusion and exclusion criteria below and then reply to this email stating “I confirm that I am eligible for this study” if it is true.

Inclusion Criteria:

- Male CrossFit Athlete
- Body Mass of equal to or greater than 70 kg
- Regular Training (1+ times/week) that involves lifting or carrying a load of at least 40 kg
- No current musculoskeletal injury (within last 3 months)

  Preferred, but not Mandatory:
  - Experience Carrying a Loaded Backpack
  - Experience Carrying Another Person using the Piggyback

Exclusion Criteria

- Failure to meet the 4 necessary inclusion criteria

Thank you very much for your interest; I look forward to working with you!

Sincerely,

Sheena Graham

INSTRUCTIONAL EMAIL - Participant

LOAD CARRIAGE STUDY - WHAT TO EXPECT

Instructions – Please Follow:

- **Please come at to the Pan Am Clinic Foundation at [time TBA] on [date TBA].**
    - **Enter the main doors of the Pan Am Clinic, turn right, and wait at the elevator across from the entrance to Diamond Athletic Supplies**
    - **A research assistant will come down the elevator to bring you up to the lab**

- **Please bring or wear form-fitting clothing, ideally of one solid colour and preferably black. Avoid green, red, beige and white.**

- **Once in the laboratory, you will have an opportunity to change into the appropriate attire and will be encouraged to ask any questions and address any concerns that you may have before signing the informed consent form. You will have had the opportunity to previously read the informed consent form as it was sent via email with this message. You will also be asked to disclose any relevant musculoskeletal injury and your previous history with carrying loads.**

Data Collection – What You Will be Doing:

- The participants will be assigned a number in the order in which they have signed in. At this time, you will also choose an envelope that will determine the order in which you will carry the loads.

- A research assistant will ask to weigh you and measure your height, and will use their hands to locate bony prominences and body landmarks so that they will know where to place specific markers. These markers will be made of green painter's tape and beige masking tape and they will either be attached to your clothing or skin, depending on the location.

- You will have an opportunity to become familiar with the walkway and the set-up. You will be asked to try to place your feet so that the right foot will come in full contact with the force plate, and only the force plate, while you walk over the runway. The goal is to learn where to start, and then start from that location while trying to walk as naturally as possible, looking forward and not down at your feet. You will walk for three successful trials with no load, and then three successful trials with each loaded condition.

- At some point after you have finished all three conditions, the researcher will let you know that you are free to change into your street clothes. Thank you very much for taking time out of your busy schedule to make this study and my thesis project possible!

***** Please be aware of the safety and welfare of the piggybacked child throughout testing. Let the researchers know of any concern you have for the child immediately. Thank you.**

INSTRUCTIONAL EMAIL - Passenger

LOAD CARRIAGE STUDY - WHAT TO EXPECT

Instructions – Please Follow:

- **Please come to the Pan Am Clinic Foundation as early as possible on Tuesday April 8, 2014.**
    - **Enter the main doors of the Pan Am Clinic, turn right, and wait at the elevator across from the entrance to Diamond Athletic Supplies**
    - **A research assistant will come down the elevator to bring you up to the lab**

- **Please bring or wear form-fitting clothing, ideally of one solid colour and preferably blue or grey. Please avoid green, beige and white (the colour of the markers).**

- **Once in the laboratory, you will have an opportunity to change into the appropriate attire and will be encouraged to ask any questions and address any concerns that you may have before the parent or guardian signs the informed consent form and the child signs the assent form. You will have had the opportunity to previously read these forms as they were sent via email with this message.**

Data Collection – What You Will be Doing:

- The passenger will stand on a scale while their height and weight are measured. The weight of the bag will be modified to equal the weight of the passenger.

- A research assistant will use their hands to locate bony prominences and body landmarks so that they will know where to place specific markers. These markers will be made of green painter's tape and beige masking tape and they will either be attached to your clothing or skin, depending on the location. Please see the consent form for the specific locations.

- You will have an opportunity to become familiar with the walkway and the set-up. The participants will be asked to try to place their feet so that the right foot will come in full contact with the force plate, and only the force plate, while walking over the runway. They will try to learn where to start, and then start from that location while trying to walk as naturally as possible. They will walk for three successful trials with no load, and then three successful trials with each loaded condition. When it is time for the piggyback condition, the carrier and passenger will need to communicate in order for the passenger to hop up onto the back of the carrier. The passenger will be asked to **please try to maintain the same position and grasp** from trial to trial and carrier to carrier.

- After all of the participants have successfully completed their piggyback trials, the researcher will let you know that you are free to change into your street clothes. Thank you very much for taking time out of your busy schedule to make this study and my thesis project possible!

Just to show you what the pilot study was like:

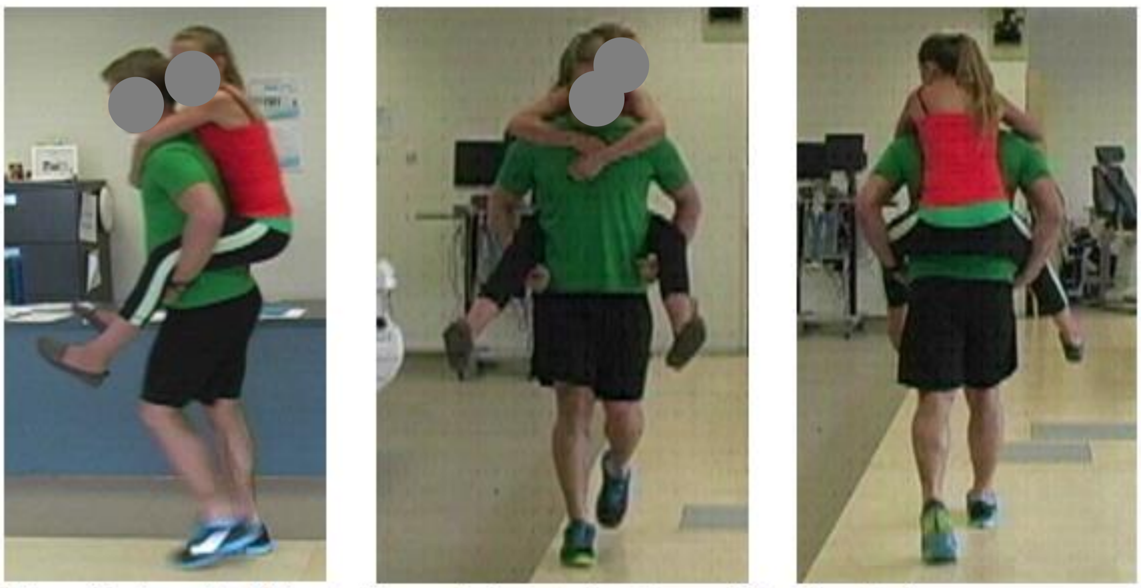

Figure 1: a) sagittal, b) anterior, and c) posterior views of the piggyback

DEBRIEF

Post-Data Collection
STATEMENT OF EXPLANATION

These videos will be analysed using Dartfish Software, and will be used to measure joint angles and gait parameters. Maximal trunk flexion, hip flexion, and knee flexion angles will be measured, as well as velocity, cadence, and stance, single stance, and double stance durations. The video frame containing maximal trunk flexion angle will be used in a lumbar spine model to calculate estimations of trunk extensor muscle torque and force, and compressive, shear and resultant lumbosacral joint reaction forces. The force plate will collect vertical ground reaction force (vGRF) data over the duration of one right foot step for each trial, and will be recorded in excel spreadsheets. The vGRFs will be graphed over time, and four specific magnitudes will be measured: the existence of an initial impact peak and its magnitude, the vGRF at the first maximal peak, the lowest vGRF at the dip between the first and second peaks, and the vGRF at the second maximal peak.

Trunk flexion is expected to increase with load, and may be greater when carrying a backpack as opposed to a person. Trunk flexion and the compressive and shear lumbosacral joint reaction forces will be the main variables considered with respect to risk to the lumbar spine. All variables will be compared between conditions using a repeated measures ANOVA test, as well as a Bonferroni test when results are significant. Both loaded conditions are expected to demonstrate kinematic differences from the unloaded condition, but it is unclear whether there will be significant differences between the piggyback and backpack conditions.

Summary: Each variable will be compared between the no load, backpack, and piggyback conditions. Any significant differences will be reported. A summary of results will be emailed to participants in approximately January 2015.

APPENDIX D

Data Tables

DATA TABLES

Kinematic Variables

Trunk	Max Flexion (°)		
P	NL	BP	PB
1	1.3	9.0	7.0
2	0.7	11.7	8.7
3	2.0	12.3	13.0
4	0.3	12.7	9.0
5	1.0	14.3	15.0
6	3.7	16.7	14.0
7	0.0	14.0	7.0
8	0.0	14.7	11.3
9	1.0	11.7	7.7
10	0.3	12.7	10.7
11	0.0	8.3	5.0
12	1.0	11.0	7.7
Avg	0.9	12.4	9.7
S.D.	1.1	2.4	3.1

Trunk	Max Extension (°)		
P	NL	BP	PB
1	-1.0	1.0	4.2
2	-0.7	8.2	7.7
3	-1.7	5.0	8.3
4	-2.0	9.0	8.0
5	0.0	11.3	12.3
6	1.0	13.0	13.0
7	-2.7	10.7	5.3
8	-1.0	10.3	8.3
9	-0.3	8.7	4.7
10	-4.0	7.7	8.7
11	-1.3	5.3	2.3
12	-1.0	8.0	5.3
Avg	-1.2	8.2	7.3
S.D.	1.3	3.2	3.2

Trunk	Flex-Ext R of M (°)		
P	NL	BP	PB
1	2.3	8.0	2.8
2	1.3	3.5	1.0
3	3.7	7.3	4.7
4	2.3	3.7	1.0
5	1.0	3.0	2.7
6	2.7	3.7	1.0
7	2.7	3.3	1.7
8	1.0	4.3	3.0
9	1.3	3.0	3.0
10	4.3	5.0	2.0
11	1.3	3.0	2.7
12	2.0	3.0	2.3
Avg	2.2	4.2	2.3
S.D.	1.1	1.7	1.1

Hip	Max Flexion (°)		
P	NL	BP	PB
1	30.4	35.7	36.8
2	31.0	36.3	35.0
3	28.7	41.0	35.0
4	26.3	39.3	34.0
5	31.3	40.0	43.0
6	22.7	38.3	36.7
7	27.0	46.0	38.0
8	29.7	41.0	35.7
9	26.7	38.7	30.7
10	22.3	34.3	34.7
11	27.0	37.3	36.0
12	27.0	32.7	30.3
Avg	27.5	38.4	35.5
S.D.	2.9	3.5	3.3

Knee	Max Flexion (°)		
P	NL	BP	PB
1	27.0	27.4	25.7
2	28.7	26.7	26.0
3	20.0	27.3	23.0
4	22.3	29.0	26.0
5	16.7	21.7	23.7
6	20.3	22.0	19.3
7	16.7	28.7	24.7
8	14.3	19.3	17.7
9	7.7	18.7	18.7
10	-4.0	20.3	19.7
11	21.3	25.7	23.3
12	22.3	23.7	23.0
Avg	17.8	24.2	22.6
S.D.	8.8	3.7	3.0

External Lumbosacral Torque Variables – Load Torques

Resultant T by Load Only (Nm)		
P	PB	BP
1	20.57	55.95
2	19.34	56.90
3	14.08	57.85
4	15.89	55.95
5	11.57	48.36
6	7.12	47.42
7	14.15	39.83
8	20.73	54.05
9	20.23	59.74
10	12.79	55.00
11	23.71	64.48
12	20.00	60.69
Avg	16.68	54.69
S.D.	4.84	6.67

Tot. Mag. T by Load Only (Nm)		
P	PB	BP
1	37.73	55.95
2	34.00	56.90
3	31.92	57.85
4	35.46	55.95
5	31.43	48.36
6	27.45	47.42
7	33.16	39.83
8	31.30	54.05
9	34.00	59.74
10	29.72	55.00
11	39.18	64.48
12	35.74	60.69
Avg	33.42	54.69
S.D.	3.33	6.67

Torque Caused by Body of Carrier (Nm)			
P	NL	PB	BP
1	-27.42	-40.26	-49.62
2	-24.04	-33.54	-36.65
3	-33.13	-40.39	-49.57
4	-18.10	-29.81	-37.85
5	-31.96	-50.01	-56.93
6	-40.01	-48.64	-62.24
7	-24.92	-37.67	-47.63
8	-31.46	-45.71	-60.97
9	-31.14	-34.75	-51.92
10	-23.04	-36.40	-46.12
11	-18.06	-25.85	-36.11
12	-28.65	-41.31	-49.14
Avg	-27.66	-38.70	-48.73
S.D.	6.42	7.27	8.75

Total Resultant Torque (Nm)			
P	NL	PB	BP
1	-27.42	-19.69	6.33
2	-24.04	-14.19	20.25
3	-33.13	-26.31	8.28
4	-18.10	-13.92	18.1
5	-31.96	-38.44	-8.57
6	-40.01	-41.51	-14.82
7	-24.92	-23.52	-7.8
8	-31.46	-24.98	-6.92
9	-31.14	-14.51	7.82
10	-23.04	-23.61	8.88
11	-18.06	-2.14	28.37
12	-28.65	-21.31	11.55
Avg	-27.66	-22.01	5.96
S.D.	6.42	10.73	13.15

Total Magnitude of Torque (Nm)			
P	NL	PB	BP
1	27.42	77.99	105.57
2	24.04	67.53	93.55
3	33.13	73.27	107.42
4	18.10	66.04	93.80
5	31.96	81.44	105.29
6	40.01	76.09	109.66
7	24.92	71.08	87.46
8	31.46	77.02	115.02
9	31.14	68.75	111.66
10	23.04	66.11	101.12
11	18.06	65.04	100.59
12	28.65	77.05	109.83
Avg	27.66	72.28	103.41
S.D.	6.42	5.58	8.34

Total Negative Torque (Nm)			
P	NL	PB	BP
1	-27.42	-48.84	-49.62
2	-24.04	-40.86	-36.65
3	-33.13	-49.79	-49.57
4	-18.10	-39.98	-37.85
5	-31.96	-59.94	-56.93
6	-40.01	-58.80	-62.24
7	-24.92	-47.30	-47.63
8	-31.46	-51.00	-60.97
9	-31.14	-41.63	-51.92
10	-23.04	-44.86	-46.12
11	-18.06	-33.59	-36.11
12	-28.65	-49.18	-49.14
Avg	-27.66	-47.15	-48.73
S.D.	6.42	7.62	8.75

Total Positive Torque (Nm)			
P	NL	PB	BP
1	0.00	29.15	55.95
2	0.00	26.67	56.9
3	0.00	23.48	57.85
4	0.00	26.06	55.95
5	0.00	21.50	48.36
6	0.00	17.29	47.42
7	0.00	23.78	39.83
8	0.00	26.02	54.05
9	0.00	27.12	59.74
10	0.00	21.25	55
11	0.00	31.45	64.48
12	0.00	27.87	60.69
Avg	0.00	25.14	54.69
S.D.	0.00	3.88	6.67

Internal Lumbosacral Torque and Force Variables

Resultant Muscle Torque (Nm)			
P	NL	PB	BP
1	27.42	19.70	-6.33
2	24.04	14.20	-20.25
3	33.13	26.31	-8.27
4	18.10	13.91	-18.10
5	31.96	38.44	8.57
6	40.01	41.52	14.82
7	24.92	23.51	7.80
8	31.46	24.98	6.92
9	31.14	14.51	-7.83
10	23.04	23.61	-8.88
11	18.06	2.14	-28.37
12	28.65	21.31	-11.56
Avg	27.66	22.01	-5.96
S.D.	6.42	10.73	13.15

	Resultant Muscle Force (N)		
P	NL	PB	BP
1	456.94	328.28	-105.52
2	400.63	236.58	-337.52
3	552.10	438.46	-137.87
4	301.67	231.88	-301.61
5	532.69	640.70	142.80
6	666.87	691.96	247.04
7	415.26	391.88	130.00
8	524.27	416.33	115.34
9	518.96	241.90	-130.42
10	384.02	393.53	-147.97
11	300.96	17.11	-474.83
12	477.45	355.13	-192.58
Avg	460.98	365.31	-99.43
S.D.	107.08	181.99	219.39

F compression			(N)
P	NL	PB	BP
1	35.27	410.71	844.61
2	-2.474	408.36	982.56
3	-83.55	276.88	853.3
4	121.29	437.87	971.458
5	7.99	146.77	644.77
6	-149.85	71.85	516.86
7	76.945	347.11	609.09
8	23.34	378.057	679.15
9	-39.45	484.4	856.82
10	41.83	279.11	820.71
11	129.5	641.56	1150.18
12	13.02	382.14	929.95
Avg	14.48842	355.4014	821.6215
S.D.	79.57441	150.2735	181.0157

F shear			(N)
P	NL	PB	BP
1	281.82	423.12	423.18
2	227.97	369.27	369.32
3	268.27	409.57	409.63
4	242.17	383.47	383.53
5	309.57	450.87	450.93
6	296.03	437.33	437.384
7	281.82	423.12	423.18
8	313.54	454.84	454.89
9	274.55	415.85	415.91
10	243.83	385.13	385.18
11	246.47	387.77	387.83
12	280.83	422.13	422.19
Avg	272.2392	413.5392	413.5962
S.D.	27.43402	27.43402	27.43459

F resultant			(N)
P	NL	PB	BP
1	285.86	594.04	945.08
2	228.49	551.32	1051.5
3	283.89	496.63	947.33
4	271.97	582.26	1045.13
5	310.06	479.75	787.1
6	338.17	446.74	648.84
7	316.82	549.65	742.86
8	318.37	600.03	818.45
9	279.44	639.19	952.59
10	249.21	480.02	907.22
11	281.54	750.32	1213.82
12	281.17	572.13	1022.02
Avg	287.0825	561.84	923.495
S.D.	30.49991	82.96703	155.2195

Vertical Ground Reaction Force Variables

Relative vGRF 1st Peak (*BW)			
P	NL	PB	BP
1	1.29	1.07	1.09
2	1.12	1.07	1.04
3	1.19	1.27	1.18
4	1.04	1.07	0.99
5	0.98	1.06	0.95
6	1.11	1.10	1.09
7	1.08	1.14	1.10
8	1.04	1.06	1.04
9	0.94	1.08	1.00
10	1.12	1.14	1.07
11	1.04	1.05	1.04
12	1.05	1.04	1.03
Avg	1.08	1.10	1.05
SD	0.09	0.06	0.06

Relative vGRF Dip (*BW)			
P	NL	PB	BP
1	0.75	0.71	0.71
2	0.69	0.69	0.70
3	0.71	0.65	0.66
4	0.66	0.69	0.66
5	0.82	0.78	0.83
6	0.72	0.72	0.70
7	0.72	0.68	0.60
8	0.70	0.67	0.68
9	0.87	0.79	0.83
10	0.80	0.70	0.68
11	0.82	0.85	0.85
12	0.76	0.76	0.75
Avg	0.75	0.72	0.72
SD	0.06	0.06	0.08

Relative vGRF 2nd Peak (*BW)			
P	NL	PB	BP
1	1.15	1.04	1.04
2	1.23	1.13	1.13
3	1.09	1.06	1.03
4	1.10	1.02	1.07
5	1.09	1.04	1.04
6	1.15	1.11	1.16
7	1.18	1.10	1.14
8	1.12	1.08	1.08
9	1.04	1.06	1.06
10	1.09	1.05	1.15
11	1.08	1.00	1.02
12	1.11	1.08	1.09
Avg	1.12	1.06	1.09
SD	0.05	0.04	0.05

vGRFAbsolute vGRF 1st Peak (N)			
P	NL	PB	BP
1	1082.20	1198.60	1225.94
2	756.37	1026.90	996.90
3	949.49	1377.39	1279.92
4	751.04	1074.71	994.64
5	903.48	1273.15	1141.32
6	978.36	1278.75	1262.63
7	906.88	1275.68	1238.39
8	969.16	1289.09	1259.32
9	765.69	1189.53	1096.35
10	808.11	1146.59	1082.34
11	764.43	1070.32	1059.76
12	873.77	1158.82	1152.32
Avg	875.75	1196.63	1149.15
SD	107.85	106.04	103.72

vGRFAbsolute vGRF Dip (N)			
P	NL	PB	BP
1	627.10	794.05	800.82
2	466.41	660.57	668.72
3	568.93	702.86	714.86
4	477.43	687.73	660.01
5	755.40	940.70	998.80
6	635.18	838.88	814.12
7	603.74	766.64	678.05
8	651.98	809.46	825.23
9	706.46	866.18	912.97
10	576.44	704.27	689.65
11	600.27	863.35	863.18
12	635.59	845.97	835.57
Avg	608.74	790.06	788.50
SD	82.52	86.61	107.84

vGRFAbsolute vGRF 2nd Peak (N)			
P	NL	PB	BP
1	963.28	1161.16	1162.39
2	833.18	1089.26	1090.55
3	867.38	1141.29	1116.40
4	792.19	1021.21	1077.76
5	998.34	1247.92	1257.28
6	1011.60	1287.03	1350.60
7	990.81	1238.63	1283.21
8	1043.45	1311.27	1312.80
9	851.63	1165.06	1164.66
10	791.79	1060.83	1164.12
11	789.76	1012.75	1036.72
12	926.47	1207.30	1223.22
Avg	904.99	1161.98	1186.64
SD	94.73	100.58	99.33

Ab.vGRF 1st Peak Diff (N)		
P	PB - NL	BP - NL
1	166.95	173.72
2	194.16	202.31
3	133.93	145.93
4	210.30	182.58
5	185.30	243.40
6	203.70	178.94
7	162.90	74.31
8	157.48	173.25
9	159.72	206.50
10	127.83	113.21
11	263.08	262.91
12	210.38	199.98
Avg	181.31	179.75
S.D.	37.83	51.68

Ab.vGRF Dip Diff (N)		
P	PB - NL	BP - NL
1	166.95	173.72
2	194.16	202.31
3	133.93	145.93
4	210.30	182.58
5	185.30	243.40
6	203.70	178.94
7	162.90	74.31
8	157.48	173.25
9	159.72	206.50
10	127.83	113.21
11	263.08	262.91
12	210.38	199.98
Avg	181.31	179.75
S.D.	37.83	51.68

Ab.vGRF 2nd Peak Diff (N)		
P	PB - NL	BP - NL
1	197.87	199.11
2	256.08	257.37
3	273.91	249.02
4	229.02	285.58
5	249.58	258.94
6	275.43	339.00
7	247.82	292.39
8	267.82	269.35
9	313.44	313.03
10	269.03	372.33
11	222.99	246.96
12	280.84	296.76
Avg	256.99	281.65
S.D.	30.48	45.92

% of Added Load that was vGRF at 1st Peak		
P	PB	BP
1	40.90	50.51
2	95.09	84.55
3	150.41	116.15
4	113.77	85.63
5	129.94	83.60
6	105.59	99.92
7	129.64	116.53
8	112.46	101.99
9	148.98	116.23
10	118.98	96.39
11	107.52	103.81
12	100.20	97.91
Avg	112.79	96.10
SD	28.70	18.62

% of Added Load that was vGRF at Dip		
P	PB	BP
1	58.68	61.06
2	68.25	71.11
3	47.08	51.30
4	73.92	64.18
5	65.13	85.56
6	71.60	62.90
7	57.26	26.12
8	55.35	60.90
9	56.14	72.59
10	44.93	39.79
11	92.47	92.42
12	73.95	70.29
Avg	63.73	63.18
SD	13.30	18.17

% of Added Load that was vGRF at 2nd Peak		
P	PB	BP
1	69.55	69.99
2	90.01	90.47
3	96.28	87.53
4	80.50	100.38
5	87.73	91.02
6	96.81	119.16
7	87.11	102.78
8	94.14	94.68
9	110.17	110.03
10	94.57	130.88
11	78.38	86.81
12	98.72	104.31
Avg	90.33	99.00
SD	10.71	16.14

Gait Variables

Walking Velocity (m/s)			
P	NL	PB	BP
1	1.36	1.22	1.18
2	1.36	1.22	1.18
3	1.24	1.28	1.19
4	1.27	1.26	1.23
5	1.07	1.17	1.05
6	1.24	1.20	1.23
7	1.25	1.29	1.33
8	1.62	1.39	1.39
9	1.24	1.27	1.23
10	1.35	1.32	1.23
11	1.31	1.25	1.55
12	1.19	1.21	1.16
Avg	1.29	1.26	1.25
SD	0.13	0.06	0.13

Cadence (steps/min)			
P	NL	PB	BP
1	113.17	108.98	104.78
2	109.50	102.78	101.34
3	104.78	100.86	101.84
4	106.86	105.80	105.77
5	101.40	110.79	105.27
6	103.72	102.28	102.31
7	101.35	103.75	106.33
8	112.40	109.04	107.92
9	97.23	97.64	96.34
10	100.41	102.78	103.28
11	100.40	99.01	92.88
12	100.39	97.20	93.81
Avg	104.30	103.41	101.82
SD	5.14	4.50	4.96

Average Step Length (m)			
P	NL	PB	BP
1	0.89	0.74	0.73
2	0.75	0.72	0.73
3	0.76	0.80	0.76
4	0.70	0.73	0.72
5	0.63	0.64	0.63
6	0.70	0.70	0.72
7	0.79	0.79	0.80
8	0.79	0.81	0.81
9	0.74	0.77	0.77
10	0.78	0.78	0.76
11	0.79	0.77	0.77
12	0.74	0.78	0.78
Avg	0.75	0.75	0.75
SD	0.06	0.05	0.05

R-R Stride Length (m)			
P	NL	PB	BP
1		1.48	1.46
2	1.51	1.44	1.42
3	1.51	1.63	1.50
4	1.37	1.44	1.41
5	1.22	1.24	1.23
6	1.32	1.36	1.39
7	1.61	1.61	1.61
8	1.56	1.61	1.60
9	1.46	1.55	1.54
10	1.44	1.57	1.54
11	1.55	1.53	1.47
12	1.43	1.56	1.57
Avg	1.45	1.51	1.48
SD	0.11	0.12	0.11

L-L Stride Length (m)			
P	NL	PB	BP
1	1.78	1.46	1.46
2	1.48	1.43	1.48
3	1.54	1.57	1.54
4	1.43	1.48	1.45
5	1.28	1.30	1.28
6	1.46	1.45	1.50
7	1.55	1.56	1.57
8	1.58	1.62	1.64
9	1.50	1.52	1.55
10	1.66	1.54	1.52
11	1.59	1.55	1.63
12	1.51	1.55	1.56
Avg	1.53	1.50	1.51
SD	0.12	0.08	0.10

R Foot Stance Time 200Hz (s)			
P	NL	PB	BP
1	0.642	0.727	0.738
2	0.690	0.780	0.775
3	0.693	0.717	0.755
4	0.707	0.740	0.755
5	0.763	0.728	0.783
6	0.737	0.777	0.787
7	0.668	0.710	0.702
8	0.687	0.712	0.732
9	0.772	0.798	0.832
10	0.722	0.712	0.735
11	0.745	0.767	0.788
12	0.747	0.793	0.813
Avg	0.714	0.747	0.766
SD	0.040	0.034	0.037

Single Stance Time (s)			
P	NL	PB	BP
1	0.41	0.36	0.37
2	0.38	0.40	0.40
3	0.41	0.43	0.41
4	0.39	0.40	0.40
5	0.41	0.37	0.37
6	0.42	0.40	0.40
7	0.53	0.40	0.40
8	0.38	0.37	0.40
9	0.44	0.43	0.43
10	0.43	0.42	0.40
11	0.43	0.40	0.43
12	0.41	0.43	0.42
Avg	0.42	0.40	0.40
SD	0.04	0.03	0.02

Double Stance Time (s)			
P	NL	PB	BP
1	0.13	0.21	0.18
2	0.17	0.19	0.18
3	0.16	0.15	0.18
4	0.19	0.18	0.20
5	0.20	0.19	0.22
6	0.18	0.20	0.21
7	0.14	0.17	0.16
8	0.16	0.17	0.18
9	0.20	0.20	0.20
10	0.17	0.18	0.19
11	0.18	0.20	0.20
12	0.21	0.20	0.22
Avg	0.17	0.19	0.19
SD	0.02	0.02	0.02

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APPENDIX F

Pilot Study

PILOT STUDY: *A BIOMECHANICAL COMPARISON OF CARRYING NO LOAD, A BACKPACK, AND A PERSON ON THE BACK*

INTRODUCTION

The purpose of this pilot study was to identify potential kinematic, kinetic, and temporal differences between walking with no load, while piggybacking a 39 kg person, and carrying a 28 kg backpack. This study also served as a practise trial for the procedures of the current study.

METHODS

Participants

Two males aged 25 and 31 with respective masses of 103 and 93 kg volunteered as participants. They both regularly strength trained with relatively heavy loads. A 39 kg 13 year old female volunteered to be the passenger for the piggyback condition. The weights and masses were compared between the force plate and the scale for validity, as seen in Table A.1. One of the filming assistants also decided to perform two of the conditions to provide extra data in case it was needed, as seen in Table A.2.

Table A.1: The weights and masses of the participant using the force plate versus the scale.

From: Object/Person	Weight (N) force plate	Mass (kg) from force plate	Mass (kg) from scale	Weight (lbs) force plate
Bag	289.7	29.53	27.4	64.97
Passenger	386.1	39.36	86.2 lbs, 39.2 kg	86.59
Participant 1	908.4	92.60	92.6	203.72
Participant 2	1018.3	103.80	103.2	228.36
(Participant 3)	807.4	82.30	81.8	181.07

Testing Protocol

The testing protocol is very similar to that of the current study. First, the participants signed an informed consent form and had markers adhered to specific landmarks: the acromion process, the anterior superior iliac spine, greater trochanter of the femur, lateral epicondyle of the femur, and lateral malleolus of the fibula. They were allowed several practise trials on the walkway in order to familiarize themselves with the location of the force plate, where one right foot step needed to land for a trial to be successful. Participant A performed the conditions in the order of no load, backpack, piggyback. Participant B performed the conditions in the order of no load, piggyback, backpack. These trials are listed in Table 10 below.

Table A.2: The testing order of conditions.

Trial	Participant 1	Participant 2	(Participant 3)
1	No Load – 1NL1	No Load –2NL1	Piggyback
2	No Load – 1 NL2	(No Load - Miss)	Piggyback
3	Backpack – 1BP1	No Load – 2NL2	Backpack
4	Backpack – 1BP2	(Piggyback - Miss)	Backpack
5	Piggyback – 1PB1	Piggyback – 2PB1	
6	Piggyback – 1PB2	Piggyback – 2PB2	
7		Backpack – 2BP1	
8		Backpack – 2 BP2	

Filming Protocol

A force plate in conjunction with a three-camera video system and motion analysis software were used to measure kinematic, gait, and ground reaction force parameters. The Advanced Mechanical Technology, Inc (AMTI) force plate that was embedded in the walkway was used to measure the ground reaction forces in three dimensions over time at a frequency of 200 Herz. The data was transferred to the research computer in the form of excel spreadsheets. Three cameras were used to film the participants, a Canon D2L camera on the right side of the walkway, a Canon D2L camera at the posterior end of the walkway, and a Canon HDV 1080i camera at the right side of the walkway that was zoomed in on the foot. All video data was recorded at 30 frames per second and was later uploaded to the Dell Inc. laptop research computer to be analyzed using 2012 Dartfish Software 6 TeamPro 6.0.10719.0.

Data Analysis

Several variables were investigated using the video data in conjunction with Dartfish Software. Trunk angles were measured by using the acromion process and greater trochanter makers for reference to draw a line representing the trunk segment, and then measuring the angle of the trunk with the vertical. Each trial was evaluated for the phase in which the greatest trunk flexion occurred and this was noted. The hip angle was measured between this line and the line represented by the thigh, which was drawn approximately from the greater trochanter of the femur to the lateral epicondyle of the femur. The knee angle was measured between an extension of the line of the thigh and the line of the lower leg, which was drawn approximately between the markers of the lateral epicondyle of the femur and the lateral malleolus of the fibula.

The ground reaction forces were graphed to illustrate their overall pattern, but further analysis was not performed because the values could not be appropriately compared between the piggyback and backpack condition due to the large difference in the magnitude of the load. The resultant muscle torque, resultant muscle force, and compressive and shear lumbosacral joint reaction forces were calculated using Dartfish body segment moment arm measurements and anthropometric data in a biomechanical lumbar spine model. Use of the same model is being proposed for the current study, and is detailed in Chapter III of the thesis proposal document.

RESULTS

In every trial of this pilot study, the maximum trunk flexion was displayed at the end of single stance phase. The average trunk flexion angle at the end of stance was 2.68 during the unloaded condition, 11.1 during the piggyback condition, and 14.7 during the backpack condition. The trunk flexion results can be seen in Table A.3. The participants may subconsciously try to maintain a more upright posture when they know they are being filmed. Participant 2 demonstrated greater trunk flexion when he was walking back to do the next trial, unaware that he was being measured at that moment in time. This phenomenon was also seen during a previous pilot study.

Table A.3: Trunk flexion angle at three frames before the end of single stance phase.

Participant.Trial #	Trunk Flexion Angle		
	No Load	39 kg Piggyback	28 kg Backpack
1.1	1	12.7	19.9
1.2	1	10.8	17.7
2.1	5.2	11.9	10.7
2.2	3.5	9	10.5
Average	2.68	11.1	14.7

The vertical ground reaction force graphs did not demonstrate anything out of the ordinary, and because the loads were not equal, it was difficult to look for differences in gait patterns other than the amount of weight that was produced by the load. One graph of each condition is provided below in Figures A.1 to A.3, where force in Newtons is graphed over frames recorded and each frame represents a time of 0.005 seconds. The unloaded condition seems to have a more distinct initial impact peak.

Because there were only two participants, the individual data has also been presented here to provide an idea of the variation between trials of the same condition. Table A.4 provides the lumbosacral load estimations, including the resultant muscle torque (Tm), the resultant muscle force (Fm), the joint compression force (Fc) and the joint shear force (Fs). These values were estimated during the end of single stance phase; values for each individual trial are on the left and the averages of these values for each participant are displayed on the right.

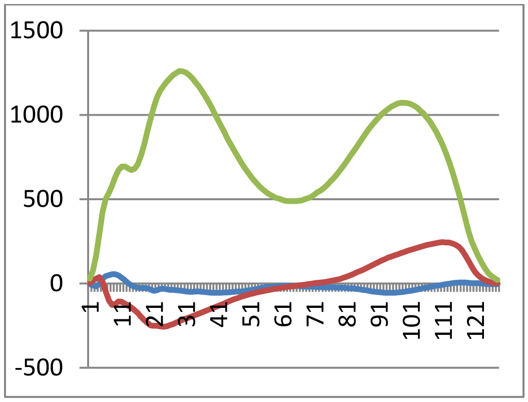


Figure A.1: The ground reaction forces during 1NL2.

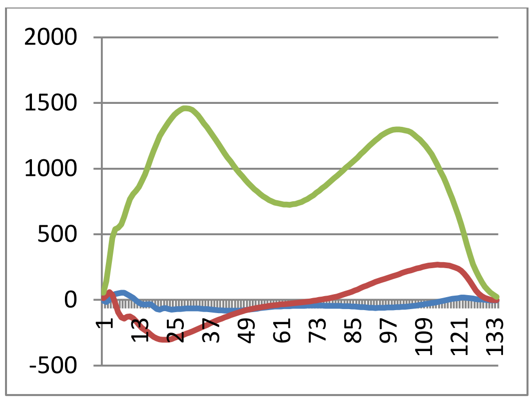


Figure A.2: The ground reaction forces during 1BP2.

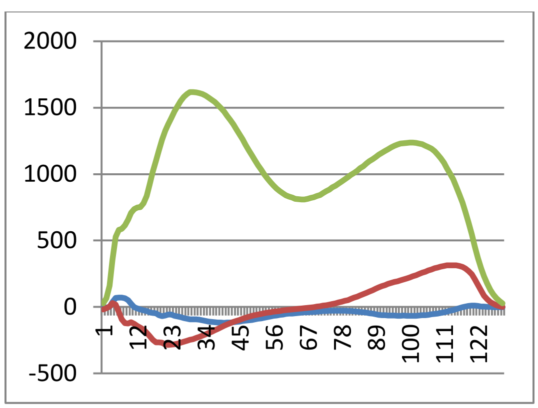


Figure A.3: The ground reaction forces during 1PB2.

Table A.4: The estimated muscle torque in Nm, muscle force in N, joint compression force in N, and joint shear force in N.

Trial	Tm (Nm)	Fm (N)	Fc (N)	Fs (N)	Mean Tm (Nm)	Mean Fm (N)	Mean Fc (N)	Mean Fs (N)
1NL1	-41	-689	1224	324				
1NL2	-36	-604	1138	324	-38.5	-646.5	1181	324
1PB1	-79	-1314	2184	522				
1PB2	-72	-1203	2071	522	-75.5	-1259	2128	522
1BP1	-39	-652	1437	467				
1BP2	-34	-562	1347	467	-36.5	-607	1392	467
2NL1	-63	-1047	1646	363				
2NL2	-40	-665	1264	363	-51.5	-856	1455	363
2PB1	-105	-1747	2681	561				
2PB2	-94	-1559	2493	561	-99.5	-1653	2587	561
2BP1	-34	-573	1423	614				
2BP2	-29	-485	1335	614	-31.5	-529	1379	614

DISCUSSION

It is unfortunate that the backpack and passenger were not of the same weight. Originally, it was planned for both loads to be the weight of the lightest person that could be found to use as the passenger. The day of the testing, it was decided that such a load seemed much heavier and potentially dangerous to carrying in a backpack. Because of this, the results are not properly comparable.

Trunk Flexion Angle

Although the piggyback load was significantly heavier than the backpack load, the backpack load produced a greater alteration in trunk flexion angle. This trend was seen more strongly in participant 1 and not strongly in participant 2. However, participant 2 also seemed to exhibit a greater trunk angle when thinking that he was not being filmed. It seems that it is important to perform a few sample measurements when the participant is walking back to the start to ensure that the trunk angle is consistent. The greatest angle of trunk flexion was seen during the end of single stance phase. In general, it seems that a greater amount of trunk flexion will be seen while piggybacking compared to backpacking with an equal load. In general, adding a posterior load produced a trunk flexion response.

Vertical Ground Reaction Forces

The vertical ground reaction forces did not exhibit any obvious differences greater than the differences produced by an additional magnitude of load. For example, every 10 N of additional load should incur an additional 10 N of vertical ground reaction force. This was observed. It is interesting to note that the unloaded condition may exhibit a more pronounced initial impact peak. It would be interesting to see if the piggybacking condition ever exhibits this as well because it seems that piggybacking may allow for a more natural gait. It would be interesting to compare the vertical ground reaction force values, at their peaks and graphed over time, between equally loaded piggyback and backpack conditions.

Lumbar Spine Model

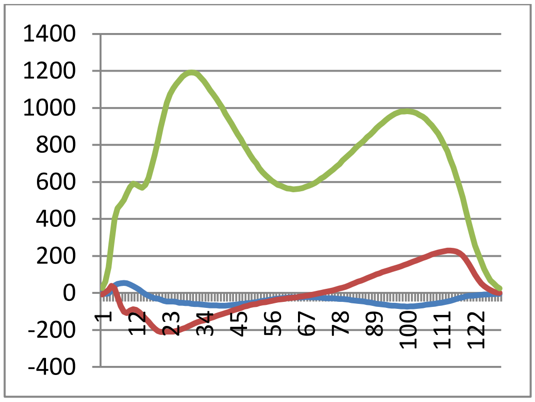
A few interesting ideas have been formed with respect to the lumbar spine model. A left lateral view camera should be added because the left arm is often behind the trunk when viewing from the right, which means that the average centre of gravity of the arms had to be estimated based on the body parts that were visible. It is interesting that the trunk flexion was greater for backpacking than piggybacking, but that every measure of lumbar load was much greater for piggybacking. The load was heavier for piggybacking, but it did not necessitate as great a change in position.

The estimated forces of joint compression were 1181/1455 N for the no load condition, 2128/2587 N for the piggyback condition, and 1392/1379 N for the backpack condition for participant 1/participant 2. For one participant, the force of joint compression was greater during backpacking than carrying no load, but it was the opposite for the other participant so it is unclear what is to be expected. It is thought that the forces may actually be higher during backpacking, because the equations only result in the resultant muscle torque. If the trunk flexors are also working much harder, which is possible, there may be a substantial amount of hidden force. The piggyback always generated the highest compressive forces, but it is unclear whether this was primarily because it involved a heavier load.

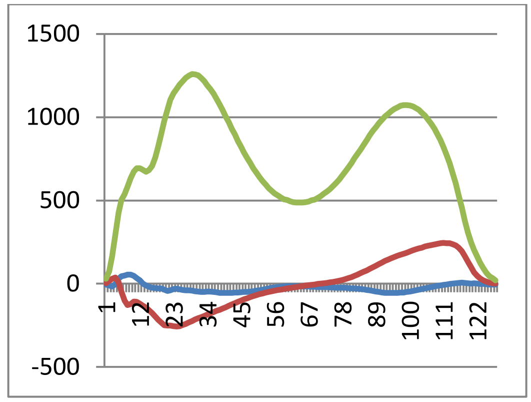
CONCLUSIONS

Differences in trunk flexion seem to exist between carrying no load, a person, and a backpack of equal weight. A larger study needs to be performed to determine if the differences are statistically significant between piggybacking and backpacking. The lumbar back model was useful in providing an idea of the biomechanics and lumbar forces during these activities. The lumbar back model should be used in conjunction with knowledge of its shortcomings. There is much to be learned about the biomechanics of piggybacking, and backpacking is a comparable activity of which there has been previous biomechanical research.

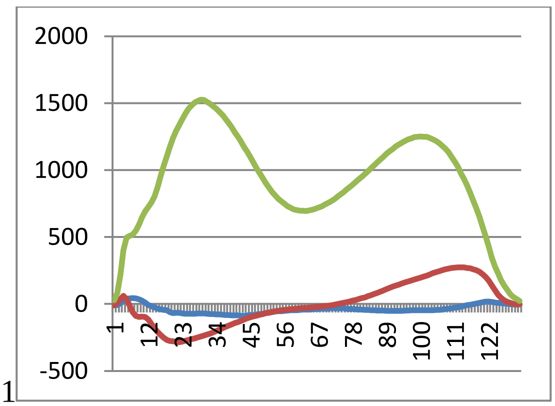
Ground Reaction Force Graphs (Participant, Condition, Trial Number)



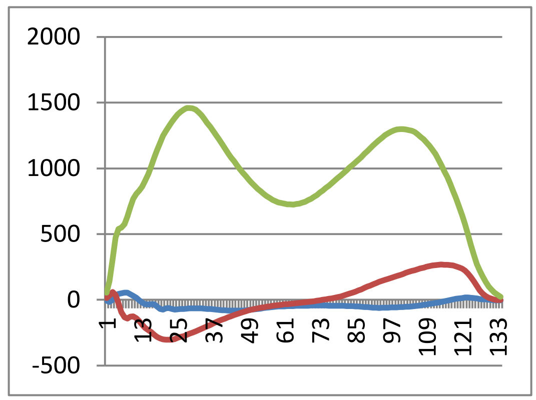
1NL1



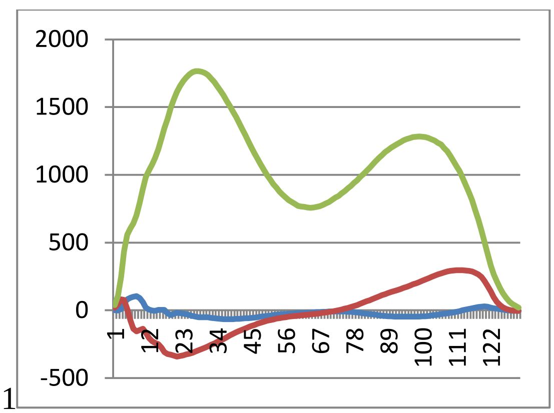
1NL2



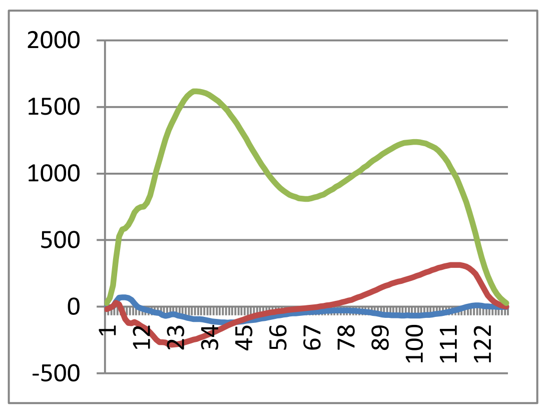
1BP1



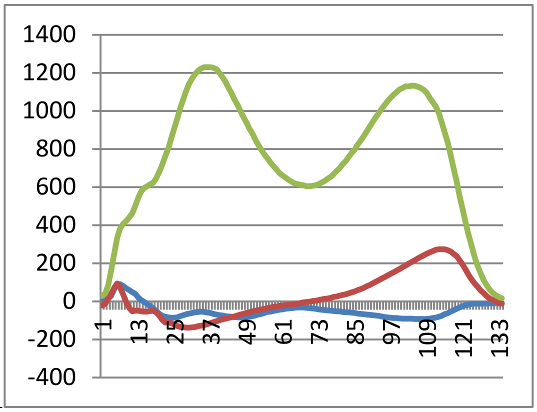
1BP2



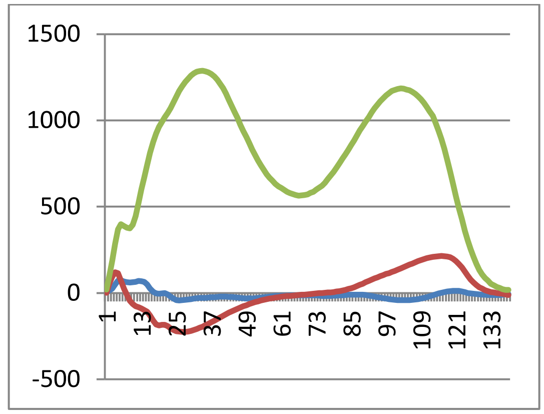
1PB1



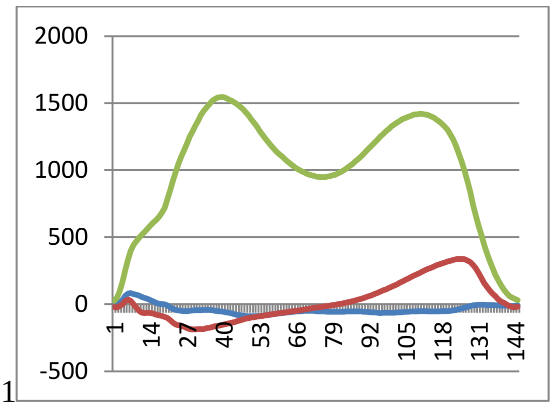
1PB2



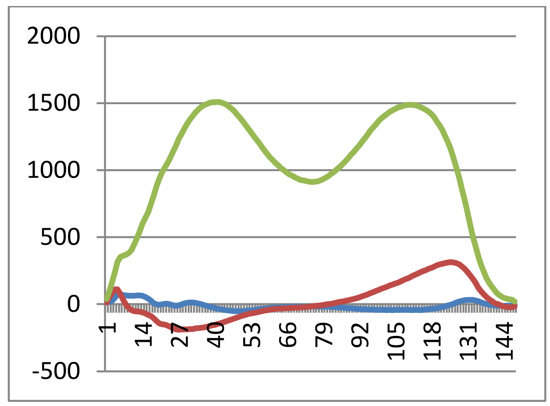
2NL1



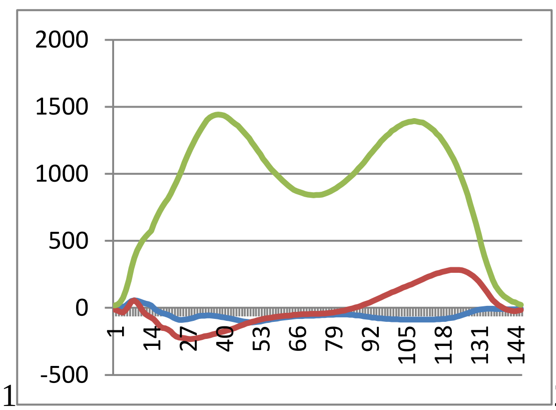
2NL2



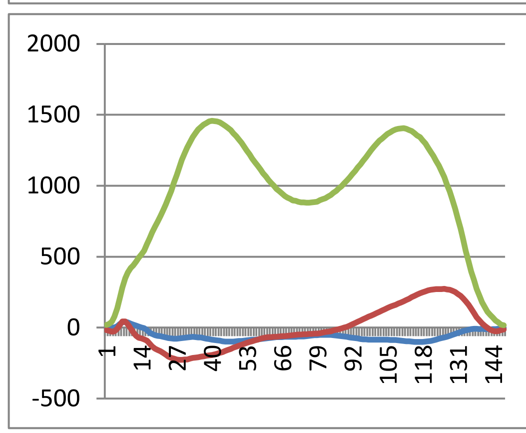
2PB1



2PB2



2BP1



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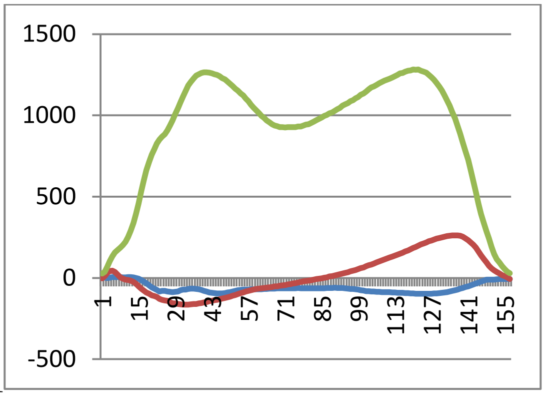
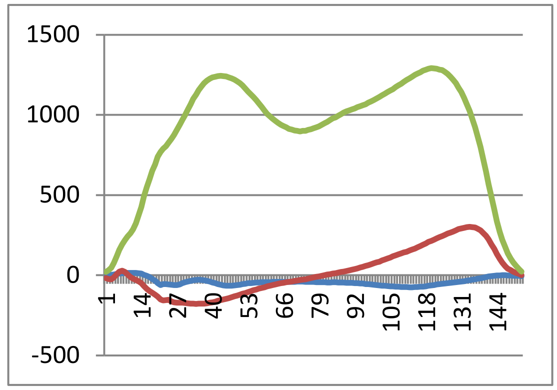
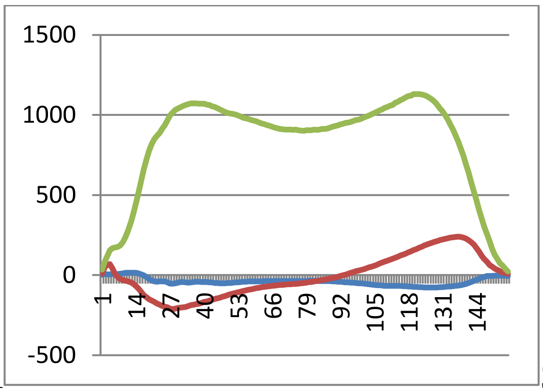


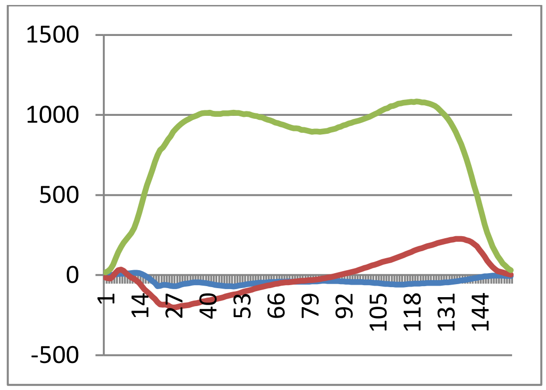
3PB1



3PB2



3BP1



3BP2