Carrying asymmetric loads during different walking conditions

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Carrying asymmetric loads during different walking conditions

by

Junsig Wang

A dissertation submitted to the graduate faculty
in partial fulfillment of the requirements for the degree of

DOCTOR OF PHILOSOPHY

Major: Kinesiology

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ABSTRACT

The purpose of this dissertation was to investigate how asymmetric load carriage affects joint kinetics and postural control during walking and stair negotiation.

In the first two studies, frontal plane joint moments were analyzed when carrying unilateral versus bilateral loads during stair negotiation. Peak L5/S1 contralateral bending moments were significantly higher when carrying a 20% body weight (BW) unilateral load for both stair ascent and descent. In addition, peak external knee varus and hip abduction moments were significantly higher in unloaded limb stance as compared to loaded limb stance. Based on our findings, we suggest that the low back and lower extremity play different roles in adjusting to asymmetric loads and recommend splitting loads bilaterally in order to decrease the frontal plane joint moments.

In the last dissertation study, we assessed postural control when carrying unilateral versus bilateral loads during walking on even and uneven surfaces. Carrying 20% BW bilateral or unilateral loads resulted in a significantly higher double stance ratio than unloaded walking. Carrying a 20% BW unilateral load or walking on an uneven surface resulted in a significantly increased coefficient of variation (CV) of double stance ratio. Unloaded limb stance had a significantly higher double stance ratio and mean medial-lateral (ML) center of pressure (COP) velocity, although the loaded limb had a significantly higher CV of ML COP velocity. We suggest that load carriage and unloaded limb stance require more conservative postural control, while asymmetric loads and uneven surfaces require more step-by-step postural adjustments.
CHAPTER 1: GENERAL INTRODUCTION AND LITERATURE REVIEW

General Introduction

Carrying loads is one of the most common tasks in activities of daily living. Individuals frequently carry heavy items on one hand instead of both hands for many reasons. Carrying loads in one hand increases frontal plane asymmetry and could be harmful to the human body due to altered postures during walking. Several studies have demonstrated that carrying asymmetric loads would result in postural and mechanical changes as compared to unloaded walking. In addition, there is evidence that asymmetric load carriage during walking increases frontal plane loading in both the low back and lower extremity.

In addition to asymmetric load carriage, individuals walk over various surfaces and often encounter challenging terrains, such as stairs and an uneven surfaces that frequently involve load carriage. Stair ascent and descent require higher knee ranges of motion and knee extension moments than level walking. Furthermore, a previous study indicated that asymmetric load carriage increases vertical ground reaction forces as compared to bilateral load carriage during stair negotiation. However, the effect of asymmetric load carriage during stair negotiation remains largely unknown, as the majority of previous research has been limited to level walking or lifting tasks. Therefore, further research should investigate the effect of asymmetric load carriage on the low back and lower extremity, addressing the potential for injury.

Changes in walking surfaces also require a higher demand for postural control. As compared to an even surface, walking on an uneven surface may result in altered postural control and gait patterns. Several studies found that an uneven surface resulted in increased step width, step time variability, medio-lateral center of mass velocity, and step variability as compared to an even surface. Also, asymmetric load carriage while walking on an uneven surface may reduce
postural stability. However, no study has evaluated postural stability of asymmetric load carriage while walking on an uneven surface. Therefore, there is a need for additional data to investigate potential concerns with risk of falls, slips, and trips when carrying asymmetric loads during walking on an uneven surface.

Literature Review

Low back pain

Low back pain is one of the most prevalent disorders and the second highest reason for visits to a physician in the United States (Deyo & Weinstein, 2001). Up to 20% of Americans will report low back pain yearly, and back pain accounts for a large proportion of health care expenditures that continue to rise without any improvements (Luo et al., 2004; Martin et al., 2009). In 1998, total health care expenditures associated with back pain in the U.S. accounted for $26.3 billion (Luo et al., 2004). Also, a study dealing with a large sample of U.S. households, the National Health Injury Survey (NHIS), indicated that back pain was a reason for about 25% of workers’ compensation costs in the United States (Guo et al., 1995). Up to one-third of patients reported that at least moderate intensity of back pain continues a year after the first occurrence of acute back pain (VonKorff & Saunders, 1996). Thus, back pain is a prevalent and chronic condition, but specific causes for back pain still remain unknown for many reasons.

There are many factors associated with increased risk of low back pain. Low back problems can begin at an early age, and the highest frequency of back pain occurs between the ages of 35 and 55 (Andersson, 1999). Musculoskeletal impairment is the most frequent reason for low back pain (Andersson, 1999). One study reported a relationship between age and risk of low back pain since older adults may have an inability to control back muscles (Iguchi et al., 2003). Another study showed evidence that increased fitness levels resulted in decreased
occurrences of low back pain (Cady et al., 1985). In addition, physical size and body mass have an influence on low back pain. A study reported that increased body mass index contributed to higher risk of a herniated disc (Heliovaara, 1987).

Due to the complex structure of the back, it is difficult to pinpoint the cause of back pain, since many interacting factors may be involved. In general, it is believed that back pain is attributed to muscle inflammation and spasms, strains, pinched nerves, sciatica, bulging disc, and alignment of spine (Whiting & Zernicke, 2008). Also, degenerative intervertebral discs and arthritis of the facet joints are the common in older adults (Buckwalter, 1995). In children, a herniated (“slipped”) disc can cause not only a severe pain, but also an increased recurrent risk and long-term risk of low back pain into adulthood (Salminen et al., 1999). A sudden vertebral migration related to degenerative discs occurs most often in the lumbosacral joint (Adams & Hutton, 1982).

Relationships between mechanical factors and low back pain have been established. The mechanical factors include lumbar lordosis, pelvic tilt, and abnormal length and strength of back muscles. Excessive lumbar lordosis has been associated with weak trunk muscles as a potential cause of low back pain (Thorstensson & Arvidson, 1982; Suzuki & Endo, 1983). Pelvis inclination also contributes to excessive lumbar lordosis and may be related to low back pain (Chaleat-Valayer et al., 2011). Moreover, weak abdominal muscles have been found in patients with low back pain (Hemborg & Moritz, 1985; Lee et al., 1995). However, other studies reported no association between pelvic tilt, lumbar lordosis, and abdominal muscle strength (Youdas et al., 1996; Youdas et al., 2000)

It is generally considered that tightness of the hamstring muscles may cause low back pain as well. Mellin (1988) reported that people with hamstring tightness exhibited an unstable
pelvic mechanism resulting in low back pain. Although there are inconsistent results, leg length discrepancy has also been related to low back pain in several studies (Nadler et al., 1998; ten Brinke et al., 1999; Rannisto et al., 2015).

**Low back models**

The lumbosacral joint (L5/S1) is the lowest lumbar region and supports the highest amount of load from the upper body. Thus, the L5/S1 is often considered the most common region that low back pain originates from. The majority of previous studies have utilized net joint moments, compressive forces, and shear forces at the L5/S1 as an indicator of low back loading (Buseck et al., 1988; Bush-Joseph et al., 1988; Cholewicki et al., 1995; Kingma & van Dieen, 2004a). Another approach for quantifying low back loading is to estimate individual muscle forces associated with L5/S1 joint motions (Cholewicki et al., 1995; Marras & Granata, 1997; Fathallah et al., 1998). Thus, many researchers have attempted to relate kinetics of the lumbosacral joint to an increased risk of low back pain.

Forces or loads imposed on the lumbosacral joint have been one of the main topics in biomechanical research estimating spinal loading related to low back pain. Many studies have proposed a variety of low back models or methods to investigate lifting tasks (Chaffin, 1969; Schultz & Andersson, 1981; Marras et al., 1984; McGill & Norman, 1986; Marras & Granata, 1997; Kingma & van Dieen, 2004b). One early low back model assumed that a single muscle vector supported the external load using a two-dimensional static analysis. This model was based on an internal force counteracting the external load to satisfy static equilibrium, but without any muscle co-contraction (Chaffin, 1969). It is generally referred to as a static sagittal plane model and thus the model has been developed to evaluate the static sagittal parameters involving compressive and shear forces at the lower lumbar spine.
During the 1980s, another model was developed to include multiple muscle forces during a lifting task. This model also attempted to predict compression forces and shear forces imposed on the spine and provided some aspects of muscle co-contraction and intra-abdominal pressure (Marras et al., 1984). In the model, EMG activity of ten trunk muscles, intra-abdominal pressure, and joint moments at the lower back were evaluated during isometric and isokinetic conditions. The outputs of the model imply differences in internal trunk mechanisms for dynamic tasks as compared to a static posture. However, this model was also two-dimensional and evaluated isokinetic trunk flexion.

Schultz and Andersson (1981) proposed a three-dimensional model including 10 trunk muscles. This model included 10 unknown forces and required a muscle recruitment optimization approach to solve the indeterminate problem. Electromyography (EMG)-assisted models were developed to compensate for the lack of muscle co-contraction of the past models. This approach involved improved anatomical modeling, estimation of muscle activations, vertebral kinematics, and consideration of muscle length, muscle velocity, and muscle cross-sectional areas.

McGill and Norman (1986) created an EMG-assisted model with 48 muscles and 7 ligaments that estimated joint moments, compressive forces, and shear forces at L4-L5. Marras & Granata (1997) developed a model to evaluate three-dimensional spine loading at the L5/S1 joint during lifting tasks. This model estimated individual muscle forces using 10 EMG channels. Kingma and van Dieen (2004b) combined a three-dimensional linked segment model with an EMG-assisted trunk model to quantify spinal loads at the L5/S1 joint during asymmetric and symmetric lifting tasks. In this model, muscle forces calculated by the EMG-driven model were added to net reaction forces at the L5/S1.
Due to the equipment required to collect input data for EMG-driven models, researchers have attempted to derive simple predictive equations for spinal loading. McGill et al. (1996) used joint moments and corresponding compressive forces from all subject trial data to create a polynomial prediction equation that predicted low back compression forces with an $R^2$ of 0.94. The value of $R^2$ (0.94) indicated that the outcomes from the regression model were very close to the original EMG-driven model. Arjmand et al. (2012) also developed a predictive equation that estimated compressive and shear forces at the L5/S1 and L4/L5 joints using four input variables: thorax flexion angle, load magnitude, load lateral position, and load sagittal position. The predictive equation was determined using a non-linear finite element model to predict spine loads for each combination of input variables during asymmetric and symmetric lifting tasks.

Another common lower back model is a Newton-Euler inverse dynamic model based on Newton’s equilibrium of forces (de Looze et al., 1992; Seay et al., 2008; Gillette et al., 2009; Faber et al., 2010). External forces and moments produced by gravity are equated to estimate the unknown muscle and joint forces. Although it is hard to quantify the force in the specific anatomical structures, this model allowed measurement of joint forces and moments during dynamic tasks such as running or walking. For example, Seay et al. (2008) used this approach to evaluate L5/S1 and T12-L1 joint moments during running at different stride lengths. In addition, Gillette et al. (2009) applied the inverse dynamics to evaluate L5S1 joint moments during carrying tasks. Thus, in spite of several limitations (e.g., no consideration of muscle co-activations, a rigid body assumption), this model has been mainly used in quantifying the lower back loading during walking or running.

As stated, there are a variety of approaches or models to estimate spinal loading, which are 2D static model, inverse dynamic method, EMG-assisted model, regression methods, and
more. In the last few decades, the majority of previous literature on low back was static lifting tasks. The EMG-assisted model has gained popularity. This approach includes consideration of muscle co-contraction and many muscle groups that contribute to force generation in the lower back model. However, this approach includes a limited number of muscle measured by the surface EMG and has been limited to lifting tasks to evaluate spinal loading. Furthermore, EMG would be normalized by comparing the static EMG and the EMG during dynamic trials. This procedure can be problematic due to unknown relationship between EMG and force generation. Also, there are the number of different tests for maximal activation (Marras et al., 2001).

Although there are some limitations of the inverse dynamics approach such as no consideration of muscle co-contraction and ligament forces, and potentially biased anthropometric data, this approach allow us to get reliable data for a variety of dynamic movements.

**Load carriage**

Carrying loads is a common activity of daily living. Individuals carry a variety of items in one hand or both hands, which may lead to asymmetrical and symmetrical load carriage. Different carrying methods can change body posture, walking patterns, muscle activity patterns, and balance during walking. Research on carrying loads has reported diverse impacts on the human body. A majority of previous research on load carriage has been related to carrying a conventional backpack. Physical stress caused by carrying a backpack alters body posture, gait kinematics, and kinetics (Martin & Nelson, 1986; Quesada et al., 2000; Chow et al., 2005; Birrell et al., 2007; Majumdar et al., 2010; Wang et al., 2013).

There is considerable evidence that carrying bilateral loads affects gait kinematics. Martin & Nelson (1986) investigated the effect of relatively heavy loads (up to 36 kg) and found that stride length and swing time decreased, while double support time and stride rate increased
as the load increased. Another study reported similar results for gait kinematics in healthy adolescent girls (Chow et al., 2005). The results showed that load carriage (up to 15% of body weight) increased double stance time and decreased walking speed. However, Chow et al. (2005) found decreased stride rate with load carriage when walking at preferred speeds (up to 1.21 m/s), which may be due to different experimental settings from the study (Martin & Nelson, 1986) that applied a constant walking speed (1.78 m/s).

Many studies have indicated that bilateral load carrying (when carrying loads on both shoulders) alters body postures and joint kinematics. An increase in forward inclination has been associated with increasing loads (Martin & Nelson, 1986; Pascoe et al., 1997; Quesada et al., 2000; Hong & Cheung, 2003). Hong and Cheung (2003) studied primary school boys age 9-10 years old while carrying 0, 10, 15, and 20% of their body weight (BW) during level walking. They found significantly greater trunk inclination when carrying a 20% BW bag compared to 0, 10, and 15% BW load conditions. Majumdar et al. (2010) reported that military loads (up to 17.5 kg) resulted in more ankle dorsiflexion, knee flexion, and hip flexion during heel strike. Birrell and Haslam (2009) studied military load carriage (up to 32 kg) and found an increase in knee flexion and pelvis tilt with increasing load. A more recent study demonstrated that load carriage (32 kg) resulted in higher pelvis tilt, hip flexion, and knee flexion during heel strike when compared to no load (Wang et al., 2013). Across studies, various adaptive patterns in lower extremity joints and the trunk during bilateral load carriage have been consistently observed.

Changes in ground reaction forces (GRF) and joint kinetics when carrying symmetric loads have been found in several studies (Chow et al., 2005; Birrell et al., 2007; Wang et al., 2013). One study indicated that increasing loads (up to 40 kg) are linearly associated with peak
anterior-posterior and vertical GRF (Birrell et al., 2007). Chow et al. (2005) studied healthy schoolgirls age 10-15 years old and reported increases in hip abduction, hip internal and external rotation, knee extension, knee valgus, and ankle plantar flexion moments during stance as load increased. In addition, they found increases in the peak anterior-posterior and the peak vertical GRF with increasing loads up to 15% of body weight. Wang et al. (2013) also found significantly higher hip extensor and knee extensor moments when carrying a 32 kg military bag compared to no load.

Other research has shown that load carriage causes higher demand on metabolic cost that may induce fatigue. Epstein et al. (1988) found significant increases in energy cost while carrying 40 kg vs. 25 kg loads for 2 hours. Hong et al. (2000) found increases in energy expenditure and oxygen uptake when carrying a 20% BW backpack compared to a 10% BW backpack in 10 year old children. To be specific, when carrying 20% BW load, the children showed 30 beats per minute higher than no load and a 4.6% VO$_2$ max increase in metabolism when compared to the no load. Lloyd and Cooke (2000) studied two different backpacks (25.6 kg) during uphill and downhill gradients on treadmill. The results of the study indicated 53% and 45% higher mean oxygen consumption (VO$_2$) during two types of backpack carriage as compared to the unloaded condition.

Asymmetrical loading can occur when load center of mass is shifted laterally (DeVita et al., 1991). This carrying method, on one shoulder or in one hand, is utilized for a variety of reasons. A study investigated the impact of carrying asymmetric loads on static posture and gait kinematics in children with no bag and while carrying an one-strap book bag, a two-strap bag, and a one-strap athletic bag (Pascoe et al., 1997). The results indicated that the load of the book bags (7.7 kg, 17% of body weight) produced lateral spinal bending with the asymmetrical
athletic bag. DeVita et al. (1991) reported that the asymmetric load caused unilateral and unbalanced use of trunk muscles on the non-loaded body side for both stance phases and proposed that asymmetric load carrying may be a greater risk factor for injury than symmetric carrying.

Fowler et al. (2006) provided evidence that asymmetric load carriage alters the kinematics of the spine and induces stature loss. They used a specific mail bag (17.5% body weight load) to simulate the task of the postal worker and found up to a 5° increase in thoracic forward flexion and up to a 12° increase in lumbar lateral bending. In addition, stature loss was shown in the loaded condition, indicating spinal shrinkage. Macias et al. (2008) reported that an asymmetric backpack caused higher shoulder pressure than a symmetric backpack carried by children 12-14 years old. Thus, asymmetric load carriage would increase postural changes in both the lower body and the upper body as compared to bilateral load carriage.

Asymmetric load carriage is also associated with physiological responses such as higher oxygen uptake and heart rate than symmetric load carriage (Ikeda et al., 2008). Legg and Cruz (2004) showed that a 6 kg asymmetric load caused more restrictive lung function than a symmetric load. Neuschwander et al. (2010) investigated the effects of asymmetric load carriage on spinal loading and spinal kinematics. They used magnetic resonance imaging (MRI) to measure that a school bag compressed lumbar disc heights in children as well as increased lateral flexion of the lumbar vertebrae. The L5/S1 disc during the 12 kg load condition was about twice as compressed as the T12–L1 disc.

In sum, previous studies have shown that asymmetrical load carrying creates a variety of issues with trunk motion, producing abnormal postures, spinal stress, and fatigue in the human body. Even though previous research examined various effects of different load conditions, little
research has been done to show exactly how carrying an asymmetric load, compared to carrying symmetric loads, affects lower back mechanisms since most studies have evaluated the lower back during static lifting tasks. Therefore, more research should be conducted to investigate the effect of asymmetric load carrying on lower back (L5/S1 joint), providing an insight into potential for injury or low back pain. The present study will be an attempt to simulate real life loading conditions utilizing two bags. Furthermore, the evaluation of carrying loads may reveal potential injurious factors regarding dynamic tasks.

Stair negotiation

Stair negotiation is a fundamental task of daily living. Many studies have reported that the biomechanics of stair waking are different from normal walking. Stair walking requires more balance adjustment than normal gait because of greater frontal trunk motion relative to pelvis (Krebs et al., 1992). Stair walking demands greater sagittal plane knee range of motion, angular velocities, and maximum knee moments when compared to normal walking (Andriacchi et al., 1980; Jevsevar et al., 1993; Rowe et al., 2000; Riener et al., 2002). Andriacchi et al. (1980) reported increased range of motion (+20°) at the hip during stair ascent. Jevsevar et al. (1993) evaluated walking, sit-to-stand, stair ascent, and stair descent and reported that stair ascent required the greatest knee range of motion and knee extension moment among the conditions. Indeed, sagittal knee range of motion was +45% higher during stair ascent (91.8° ± 10.4°) than normal walking (63.3° ± 8.1°). Also, increased knee extension moments were found during stair descent (11.9 ± 2.9 Nm/%BW) as compared to normal walking (7.4 ± 2.9 Nm/%BW).

Similarly, Riener et al. (2002) reported increases in vertical GRF and knee extension moments, but decreases in anterior-posterior GRFs during stair ascent. They also found greater
hip and knee ranges of motion during stair ascent than level walking, but smaller hip ROM during stair descent. Costigan et al. (2002) found that hip and knee shear forces were higher during stair ascent than during level walking. Surprisingly, the patella-femoral contact force was eight times higher during stair ascent as compared to level walking. It was revealed that patella-femoral forces were about 2.1 times body weight when knee flexion angle was at 60 degrees (Perry et al., 1975). Thus, higher demand of knee flexion during stair descent can result in much greater patella-femoral forces while increasing activation of quadriceps muscle group and a net quadriceps moment.

Stacoff et al. (2005) investigated vertical GRF patterns during stair negotiation. The results indicated that the vertical GRF during stair descent were considerably different than level walking. The first peak value of the vertical GRF curve (1.56 N/BW) and the loading rate (12.79 BW/s) at touchdown during stair descent were significantly increased compared to those of level walking (1.19 N/BW & 7.92 BW/s). Variability of the first peak and second peak vertical GRF increased for stair ascent and descent compared to level walking. Indeed, the mean coefficient of variation for the 1\textsuperscript{st} and 2\textsuperscript{nd} peak vertical GRF increased from level walking (about 2-5%) to stair ascent and descent (5-10%). Also, asymmetry of two consecutive steps for vertical GRF parameters also increased for stair ascent compared to walking. Thus, these changes may show increased gait variability and asymmetry during stair ambulation compared to level walking.

Several studies have reported differences between stair ascent and stair descent. Greater hip and knee flexion angles were found during stair ascent, while greater ankle dorsiflexion angles were observed during stair descent (Andriacchi et al., 1980; Riener et al., 2002; Protopapadaki et al., 2007; Samuel et al., 2011). In addition, a greater frontal trunk motion relative to the pelvis was found during stair ascent than stair descent (Krebs et al., 1992). The
first peak of the vertical GRF at heel strike was higher during stair descent than stair ascent (Riener et al., 2002; Protopapadaki et al., 2007). There is evidence that the hip joint produced mostly a flexion moment during stair descent, whereas an extension moment was observed during stair ascent (Riener et al., 2002). During descent, the second peak of the knee extension moment was obviously observed, while during ascent the second peak of the moment was small and negligible (Riener et al., 2002). Also, ankle plantarflexion moments were similar during stair ascent and descent. During stair ascent, all joints were responsible for transfer of kinetic energy into potential energy, whereas during stair descent, the joint powers were negative and the potential energy was transferred to kinetic energy (McFadyen & Winter, 1988; Riener et al., 2002; Spanjaard et al., 2007; Novak et al., 2011).

McFadyen and Winter (1988) reported greater muscle activity of the vastus lateralis, semitendinosus, gluteus maximus, medial gastrocnemius, soleus, and tibialis anterior during stair ascent than during stair descent, but without statistical comparisons. Samuel et al. (2011) found that functional demands on knee extensors and flexors were higher during stair descent than stair ascent. The functional demands were estimated by the ratio of external moment during tasks divided by the muscle moments available at that joint. Spanjaard et al. (2007) focused on the behavior of the medial gastrocnemius muscle during stair negotiation. The gastrocnemius fascicle length was almost constant during push-off, indicating other muscles need to shorten to elevate the body. During stair descent, the gastrocnemius muscle fascicles were active around heel strike and contracted concentrically, which was different than was expected for energy absorption. Accordingly, the gastrocnemius tendon is stretched and then stores energy, while other muscles such as the soleus and the knee extensors act eccentrically to decelerate the body. Lee and Chou (2007) reported changes in postural control using body center of mass (COM) and
center of pressure (COP) during stair negotiation. The results indicated that the sagittal COM-COP inclination angle was increased during stair descent when compared to a transition phase (stair-to-floor). Also, they found larger and faster COM sway in the frontal plane during stair descent than stair ascent, indicating higher demand on balance control.

There are two studies on load carriage during stair negotiation, which were the focus of the first and second dissertation studies. Spanjaard et al. (2007) studied the effect of carrying 20% BW load during stair descent and found an increase in knee extension moment. Interestingly, the ankle joint power and the behavior of the medial gastrocnemius were similar to those of the no load condition. Hall et al. (2013) also investigated carrying 20% BW load during stair ambulation and focused on medial knee joint loading. They found that the peak external knee adduction moment was increased during stair ascent compared to normal walking and stair descent. Also, carrying 20% BW load resulted in higher external knee adduction moment than no load. Thus, additional load carriage during stair negotiation may result in higher knee loading.

In sum, the findings of previous studies suggest that stair ascent and descent require higher mechanical demands than walking. However, fewer studies have investigated load carriage during stair negotiation and thus the effects of asymmetric load carriage during stair negotiation remain largely unknown. Therefore, it would be valuable to investigate how asymmetric load carriage during stair negotiation affects the human body to address potential for injury.

**Stability and load carriage**

There are many studies that have attempted to define and measure gait stability related to risk of falling. The main challenge is to identify internal sources (e.g., neuromuscular) and
external sources (e.g., external loads and/or uneven surfaces) related to risk of falling (Bruijn et al., 2013). Despite substantial efforts in the field of biomechanics, there is not a consistent standard to measure dynamic stability during locomotion. Several methods have been proposed for estimating gait stability, but each has its own advantages and disadvantages.

There are a few studies focused on gait stability during load carriage. Two studies have investigated the effect of carrying loads on stability determined by maximum finite-time Lyapunov exponents (Liu & Lockhart, 2013; Graham et al., 2015). The maximum Lyapunov exponent indicates the average logarithmic rate of divergence of a system (Rosenstein et al., 1993). To be specific, the calculation of the maximum Lyapunov exponent is accomplished by identifying the nearest neighboring point in state space for each data point and then the log of the Euclidean distance curve is calculated for all neighboring points (Rosenstein et al., 1993; Bruijn et al., 2013; Liu & Lockhart, 2013; Graham et al., 2015). Finally, the slope of the curves is calculated as the divergence exponent.

Liu and Lockhart (2013) used a tri-axial accelerometer during treadmill walking to investigate the effect of carrying a load (12.7 kg) on gait stability. The local stability was quantified by the maximum Lyapunov exponent. They found that the local dynamic stability (↑maximum Lyapunov exponent) was decreased during the load condition compared to the no load condition. Recently, Graham et al. (2015) studied five load carriage conditions depending on the location of load and with or without an assistive device. Specifically, they evaluated maximum Lyapunov exponents of the segment angles (foot, shank, thigh, and pelvis) and intersegment coordination variability (shank-foot, thigh-shank, and pelvis-thigh) under the 20% BW load conditions: 1) unassisted anterior load carriage, 2) unassisted posterior carriage, 3) assisted anterior carriage, 4) anterior load carriage, and 5) unloaded gait. The results indicated
that load carriage resulted in increased coordination variability and decreased local dynamic stability (↑maximum Lyapunov exponent) as compared to unloaded gait. However, these two studies evaluated the local dynamic stability of a segment or the low back region and thus the results showed limited information about whole body stability during load carriage. Also, it is unknown whether the measure is a valid predictor of risk of falling.

Postural stability has been used as a predictor of falls. In general, stability can be defined as an ability to maintain the center of mass (COM) position within the base of support (BOS). In postural control research, COP signals were usually used to exhibit postural stability since the COP exhibits characteristics of the COM excursion as well as aspects of equilibrium control (Winter et al., 1996; Blaszczyk & Klonowski, 2001). In this sense, several previous studies on asymmetric load carriage reported COP parameters to evaluate postural stability (Zultowski & Aruin, 2008; Bampouras & Dewhurst, 2016; Vieira et al., 2016). These studies indicated that an asymmetric load with a sidepack or a briefcase resulted in higher medio-lateral (ML) COP velocity during quite standing (Zultowski & Aruin, 2008) and greater ML center of pressure (COP) displacement during gait initiation. (Vieira et al., 2016). However, a recent study reported no differences between bilateral and unilateral shopping bags in ML COP excursion and velocity during quite standing (Bampouras & Dewhurst, 2016). Thus, there is a discrepancy of previous findings and limited information of how asymmetric load carriage affects postural stability during a dynamic task such as walking.

Another useful approach is time-to-contact (TTC) analyses based on the COP or the COM to assess postural control. A main strength of this approach is that it includes both spatial and temporal (velocity and acceleration) aspects of postural control relative to the base of support (Haddad et al., 2006). Specially, TTC indicates the estimated time it takes the COP or
the COM to reach a two-dimensional boundary of the base of support. TTC of the COP has mostly been used to evaluate postural stability in quiet standing (Slobounov et al., 1998; Hasson et al., 2008). However, there are limitations when applying the TTC to gait since the COP must leave the boundary of one foot and shift to the other foot as human body progresses forward. A number of studies have used a similar TTC approach of the COM, called ‘margin of stability’, to assess stability during walking (Hof, 2008; Hak et al., 2012; McAndrew Young & Dingwell, 2012). A main assumption of this model is the inverted pendulum to modify the COM, called ‘the extrapolated COM’, during walking. The extrapolated COM (XCOM) can be calculated as:

\[ XCOM = COM + \frac{v\text{COM}}{\omega_0} \]

The inverted pendulum’s eigenfrequency is used to estimate the extrapolated COM:

\[ \omega_0 = \sqrt{\frac{g}{l}} \]

where \( g \) indicates the acceleration of gravity (9.81 m/s\(^2\)) and \( l \) is the pendulum length of the subject. The margin of stability is calculated as:

\[ b = BOS - XCOM \]

To quantify the time before the XCOM reach the base of support, \( b_t \) can be calculated as

\[ b_t = \frac{b}{v\text{COM}} \]

However, this concept may be limited to how accurately the COM can be estimated or how well the assumptions of the inverted pendulum is supported during walking. In fact, there is some doubt about for assessing postural stability during stair negotiation because of the assumption of the inverted pendulum.

Again, there is limited information about the effect of load carriage on postural stability. Postural stability during load carriage has not been investigated during stair ambulation and
walking on an uneven surface. Most previous research on TTC has investigated quiet standing, but not gait. Therefore, in the third dissertation study, we will estimate postural stability during walking on an uneven surface using TTC of the COP approach to estimate the margin of safety while carrying asymmetric loads. The TTC of the COP procedure will be described in more detail in the methods of the third dissertation study.

**Different surfaces**

Individuals walk over various surface conditions and often encounter challenging terrains such as uneven, slippery, or ramp surfaces. Several studies have shown adaptive patterns in human gait during various surface conditions. In an attempt to study the contribution of head and pelvis control in the postural control system while walking on uneven surfaces, Menz *et al.* (2003) found that a stable head played an important role in maintaining balance during walking in response to unexpected perturbations created by irregular surfaces. They also found that people maintained their walking speed but exhibited reduced cadence, increased variable step timing, and greater harmonic ratio during uneven surface walking compared to level surface walking. Thies *et al.* (2005) indicated that healthy young adults showed greater variability in step width and step time as well as greater step width during irregular surface walking. They also found that this effect of an irregular surface on gait variability was significant for healthy older adult women. For example, in spite of increased step width and decreased walking speed for the irregular surface, the older adult healthy women produced increased step width variability (+28%) and step time variability (+51%) as compared to the regular surface condition.

Gates *et al.* (2012) investigated walking on a rock surface at four different speeds. The results of this study indicated increased variability of step width and stride length during the rock surface compared to the even surface. Also, they found that walking on the rock surface resulted
in a flatter foot motion and increased knee and hip flexion, which may be responsible for lowering the COM vertically. The knee (+8°) and hip flexion (+6.9°) angles were increased for the rocky surface during the early stance phase. Therefore, these findings provide evidence that different surface characteristics induce changes in gait kinematics and variability that may be related to gait stability.

Other studies have provided evidence how stability is affected by challenging surface conditions during gait. MacLellan and Patla (2006) found that the vertical COM decreased on the compliant surface during walking, while mediolateral COM was not changed. The compliant walking surface demanded increased gastrocnemius and soleus activity during the push-off phase, which accounts for greater step length. Furthermore, although medial-lateral margin of stability was constant, anterior-posterior margin of stability was decreased while stepping on the compliant surface. In response to the compliant walking surface, the step length and step width can be increased to secure a larger base of support for control of the COM. Marigold & Patla (2008) also demonstrated greater step, trunk, and head variability while walking on a multi-surface terrain that consisted of solid, rocky, compliant, irregular, tilted, and slippery surfaces. These results may indicate that walking on the variable surface condition leads to a challenge in maintaining balance and stability.

Previous work has shown that uneven walking surfaces result in kinematic changes and increases in variability. Even though previous work has examined altered body postures under different walking surfaces, little research has been done to directly assess how walking on an uneven surface affects postural stability. Therefore, in the third dissertation study, we investigated the effect of walking on an uneven surface on postural stability.
References


CHAPTER 2: CARRYING ASYMMETRIC LOADS DURING STAIR NEGOTIATION

Abstract

Individuals often carry items in one hand instead of both hands during activities of daily living. The combined effects of carrying asymmetric loads and stair negotiation may create even higher demands on the low back and lower extremity. The purpose of this study was to investigate the effect of symmetric and asymmetric loading conditions on L5/S1 and lower extremity moments during stair negotiation. Twenty-two college students performed stair ascent and stair descent on a three-step staircase (step height 18.5 cm, tread depth 29.5 cm) at preferred pace under five load conditions: no load, 10% body weight (BW) unilateral load, 20% BW unilateral load, 10% BW bilateral load, and 20% BW bilateral load. Video cameras and force platforms were used to collect kinematic and kinetic data. Inverse dynamics was used to calculate frontal plane moments for the L5/S1 and lower extremity. A 20% BW unilateral load resulted in significantly higher peak L5/S1 lateral bending, hip abduction, and external knee varus moments than nearly all other loading conditions during stair ascent and stair descent. Therefore, we suggest potential benefits when carrying symmetrical loads as compared to an asymmetric load in order to decrease the frontal joint moments, particularly at 20% BW load.
Introduction

Individuals often carry items in one hand instead of both hands during activities of daily living such as walking and stair negotiation. Holding an object with one hand is frequently utilized when the carried object has a handle or to allow the opposite hand to be free for other activities. Previous studies have shown that level walking while carrying asymmetric loads with one strap backpacks or mailbags resulted in higher trunk lateral bending (Pascoe et al., 1997; Fowler et al., 2006), higher trunk forward lean (Fowler et al., 2006), and higher levels of perceived low back pain (Macias et al., 2008) than unloaded walking. In addition, studies have shown that walking while carrying asymmetric loads in a bag or sidepack resulted in higher hip abduction moments (DeVita et al., 1991; Matsuo et al., 2008) and higher L5/S1 bending moments (DeVita et al., 1991) than unloaded walking. These studies provide evidence that asymmetric load carriage during walking increases frontal plane loading in both the low back and lower extremity. Therefore, it is important that further research is conducted to investigate the effect of asymmetric load carriage on the low back and lower extremity in an effort to reduce the potential for injury.

Stair negotiation is an activity of daily living that commonly involves load carriage. Previous studies have reported that unloaded stair ascent and descent required higher ankle dorsiflexion angles (Riener et al., 2002), knee flexion angles (Jevsevar et al., 1993; Riener et al., 2002), and knee extension moments (Jevsevar et al., 1993; Riener et al., 2002) as compared to level walking. Hall et al. (2013) found that carrying symmetric loads of 13.6 kg (approximately 20% body weight) in a container in front of the body or in a backpack resulted in higher external knee varus moments than when carrying no load. Furthermore, stair ascent resulted in higher external knee varus moments than walking or stair descent across loading conditions. These
findings indicate that stair negotiation involves higher knee extension moments than walking (Jevsevar et al., 1993; Riener et al., 2002) and that load carriage during stair ascent may also result in higher external knee varus moments (Jevsevar et al., 1993; Riener et al., 2002; Hall et al., 2013).

The effects of asymmetrical load carriage during stair negotiation remain largely unknown, as previous asymmetrical load carriage studies have primarily focused on level walking or lifting tasks. Hong and Li (2005) found that carrying asymmetric loads in a one-strap athletic bag resulted in higher normalized vertical ground reaction forces at 10% of body weight for stair ascent and at 15% of body weight for stair descent as compared to no load (Hong & Li, 2005). These results indicate that load amount likely plays an important role in asymmetric load carriage during stair negotiation. However, few studies have been done to investigate adaptive joint mechanisms in the lower extremity and low back when carrying asymmetric loads during stair negotiation. Thus, there is a need for additional joint moment data that may provide insight for potential risk and development of lower extremity injuries and low back disorders.

This purpose of this study was to assess low back and lower extremity moments when carrying symmetric loads and asymmetric loads at several load amounts during stair ascent and stair descent. We hypothesized that 1) peak L5/S1 lateral bending moments would be significantly higher during unilateral load carriage when compared to bilateral load carriage and 2) peak hip abduction and external knee varus moments would be significantly higher during unilateral load carriage when compared to bilateral load carriage. Increases of these parameters may be associated with potential concerns of intervertebral disc strain and/or degeneration (Schmidt et al., 2007) and development of knee and hip osteoarthritis (Baliunas et al., 2002; Royer & Wasilewski, 2006).
Methods

Twenty-two healthy young adults with an age range of 20 to 36 (11 males and 11 females; age 24.2±4.3 years; height 170.8±7.7 cm; mass 67.8±13.8 kg) participated in this study. Participants were free of any pathology that would prevent them from being able to carry a 20% body weight load. Individuals were excluded if they had back, neck, leg, foot, or arm pain. Prior to participating in the study, each subject read and signed an informed consent form approved by the university’s institutional review board.

Five load conditions were tested: no load, 10% body weight (BW) bilateral load, 20% BW bilateral load, 10% BW unilateral load, and 20% BW unilateral load (Figure 1). Loads were evenly split between the right and left hands during the bilateral load conditions. Hand-held bags were filled with sealed bags of lead shot to match the four loaded conditions. The unilateral load was carried in the participant’s dominant hand. Since all participants were right-handed, they carried the hand-held bag in the right hand during the unilateral load condition. The weight carried in the bags was normalized according to each subject’s body weight. These normalized loads were based on previous studies that indicated significant kinematic and/or kinetic changes when carrying loads ranging from 10% to 20% BW (DeVita et al., 1991; Chow et al., 2005; Fowler et al., 2006; Hall et al., 2013). Participants were instructed to ascend and descend a three-step staircase (step height 18.5 cm, tread depth 29.5 cm) at a preferred pace for each condition. The order of the conditions was randomized, and each condition was repeated three times. Participants were instructed to initiate stair negotiation by using the left leg on the first step and then the right leg on the second step.

A motion analysis system with 8 high-resolution cameras (Vicon Nexus, Los Angeles, CA) was used to collect three-dimensional kinematic data during each testing condition. The
dynamic marker set included bilateral great toe, lateral mid-foot, lateral malleolus, anterior calf, lateral calf, lateral knee joint line, anterior thigh, lateral thigh, greater trochanter, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), and acromion process markers. Single sacrum and cervical markers were also included. Six additional markers (bilateral heel, medial malleolus, and medial knee joint line markers) were recreated using transformations determined from a static standing trial. Portable force platforms (AMTI, Watertown, MA) on steps one and two were used to collect ground reaction force data.

Data processing

Kinematic data were captured at 160 Hz, and noise was reduced with a fourth-order, low-pass Butterworth filter at a cutoff frequency of 6 Hz. A static trial was used to estimate joint center locations which were assumed to be stationary in the segmental coordinate systems. Kinetic data were sampled at 1600 Hz and filtered at a cutoff frequency of 6 Hz. The force data were downsampling so that kinetic and kinematic data both had corresponding data points. Segment masses, center of mass locations, and moments of inertia were obtained according to De Leva’s anthropometric model (de Leva, 1996). L5/S1 lateral bending moments and lower extremity (ankle, knee, and hip) frontal plane joint moments were calculated using inverse dynamics and rigid body assumptions. The location of the L5/S1 joint center was defined by creating a virtual point 34% of the distance from the sacrum marker to the midpoint of the ASIS markers (de Looze et al., 1992; Gillette et al., 2009).

L5/S1 lateral bending moments were analyzed during single limb stance of the first and second stair steps. In order to calculate L5/S1 lateral bending moments during double limb stance, both left and right hip kinetics would be required. However, the hip kinetics for the lead and trial leg were not available at the top of the staircase (the third stair) because of the limited
number of the force plates. Thus, single limb stance was utilized for L5/S1 lateral bending moments because the lead and trial legs were not always positioned on force platforms during double limb stance. Hip and knee frontal plane moments were analyzed during the entire stance phase of the first (left leg) and second (right leg) stair steps. Joint moments were transformed to the inferior segment coordinate axes and reported as internal joint moments with the exception of knee varus moments, which were reported as external joint moments. Peak joint moments were determined across two steps and normalized by body mass. Absolute values of peak L5/S1 lateral bending moments were analyzed to avoid cancellation of left and right bending moments. All calculations were performed using a custom Matlab code.

Statistical analyses were performed using the SPSS statistical package (version 21; SPSS Inc., Chicago, IL, USA). The effect of the different loading conditions on peak joint moments was analyzed by using repeated measures Analyses of Variance (ANOVA). A one factor ANOVA design was used, and there were 5 levels of conditions (5 load conditions). When significant main effects were found, Bonferroni post-hoc tests were performed. The level of statistical significance for all tests was set at $p < 0.05$. To test the hypotheses, pairwise comparisons included differences between the five loading conditions.

Results

Peak L5/S1 lateral bending moments

There were significant differences in peak L5/S1 lateral bending moments as a function of load condition (Table 1). L5/S1 lateral bending moments were higher when comparing a 20% BW unilateral load to all other loading conditions during stair ascent and descent ($p < 0.001$). In addition, L5/S1 lateral bending moments were higher when comparing a 10% BW unilateral load
to no load during stair ascent and descent \((p \leq 0.031)\). Ensemble curves of L5/S1 bending moments are displayed in Figure 2.

**Peak lower extremity frontal plane moments**

There were significant differences in peak hip abduction moments as a function of load condition (Table 1). Hip abduction moments were higher when comparing a 20% BW unilateral load to all other loading conditions during stair ascent and descent \((p < 0.001)\). In addition, hip abduction moments were higher when comparing a 20% BW bilateral load to no load and a 10% BW unilateral load during stair ascent \((p < 0.001)\). Hip abduction moments were also higher when comparing a 10% BW unilateral load to no load and a 10% BW bilateral load during descent \((p \leq 0.001)\), when comparing a 20% BW bilateral load to no load during stair descent \((p < 0.001)\), and when comparing a 10% BW bilateral load to no load during stair ascent and descent \((p \leq 0.022)\). Ensemble curves of hip abduction moments are displayed in Figure 3.

There were significant differences in peak external knee varus moments as a function of load condition (Table 1). Knee varus moments were higher when comparing a 20% BW unilateral load to all other loading conditions during stair ascent and descent \((p < 0.001)\). In addition, knee varus moments were higher when comparing 20% BW bilateral and 10% BW unilateral loads to no load and a 10% BW bilateral load during stair ascent and descent \((p \leq 0.001)\). Knee varus moments were also higher when comparing a 10% BW bilateral load to no load during stair ascent \((p = 0.009)\). Ensemble curves of external knee varus moments are displayed in Figure 4.

**Discussion**

The purpose of this study was to investigate the effects of carrying asymmetric loads on low back and lower extremity frontal plane moments during stair negotiation. Peak L5/S1 lateral
bending, hip abduction, and external knee varus moments were significantly dependent upon load condition. Changes in low back and lower extremity frontal plane moments when carrying asymmetric loads during stair negation may provide important preliminary knowledge concerning potential risk of injury.

**L5/S1 lateral bending moments**

The first hypothesis that peak L5/S1 lateral bending moments would be significantly higher when carrying unilateral compared to bilateral loads was partially supported. L5/S1 lateral bending moments were higher when comparing 20% BW unilateral to 20% BW bilateral loads during stair ascent and descent, but there were no differences when comparing 10% BW unilateral and bilateral loads (Table 1). Similarly, Gillette et al. (2009) reported higher L5/S1 lateral bending moments when comparing 20% BW unilateral to 20% BW bilateral loads during walking in children. Devita et al. (1991) also reported higher L5/S1 lateral bending moments when comparing a 20% BW asymmetric load to no load during walking. Thus, it appears that a 20% BW load is at or beyond a critical level where asymmetry results in a substantial increase in low back moments for both walking and stair negotiation.

It should be stressed that peak L5/S1 lateral bending moments were dramatically increased when carrying a 20% BW unilateral load during both stair ascent and descent. During stair ascent, L5/S1 lateral bending moments for a 20% BW unilateral load were 72% higher than no load and 54% higher than a 20% BW bilateral load (Table 1, Figure 2). Furthermore, during stair descent, L5/S1 lateral bending moments for a 20% BW unilateral load were 75% higher than no load and 50% higher than a 20% BW bilateral load. These large increases in L5/S1 lateral bending moments may indicate an increased risk of low back injury. Increased L5/S1 lateral bending moments are linked to increased compressive spinal loading, lateral shear
loading, and trunk muscle co-contraction (Marras & Granata, 1997). A finite element study demonstrated that lateral bending moments resulted in shear strains in the annulus fibrosus (Schmidt et al., 2007). Further, lateral bending moments combined with axial rotation moments can lead to rupture in the disc fibers (Schmidt et al., 2007). Thus, greater than 20% BW asymmetric load carriage may result in substantial low back loading and be potentially injurious.

Another interesting finding was that the L5/S1 lateral bending moments for a 20% BW unilateral load were directed toward the left side of the body (Figure 2). A left lateral bending moment (positive values) was toward the opposite side of the body (contralateral bending) where the unilateral load was carried. Furthermore, peak L5/S1 lateral bending moments for a 20% BW unilateral load occurred during step two of stair ascent and descent. Participants contacted step two with their right leg, which is the same side as the carried load during the unilateral conditions. Thus, it appears that the lower back is exposed to the highest lateral bending moments when the leg on the loaded side is performing a step during unilateral load carriage on stairs. However, further tests that alternate the lead leg or use a staircase with more steps are needed to rule out potential differences in loading between steps one and two.

**Lower extremity frontal plane moments**

The second hypothesis that hip abduction and external knee varus moments would be higher when comparing unilateral to bilateral load carriage was partially supported. Hip abduction and knee varus moments were higher when comparing 20% BW unilateral to bilateral loads during stair ascent and descent. Knee varus moments were also higher when comparing 20% BW unilateral to bilateral loads during stair ascent and descent, but hip abduction moments were only higher during stair descent (Table 1). As with the low back, it appears that a 20% BW load is at or beyond a critical level where asymmetry results in increases for hip and knee
moments. Previous studies reported increased hip abduction and knee varus moments when carrying bilateral loads during walking and stair negotiation (Chow et al., 2005; Hall et al., 2013). Our findings demonstrate that increases in hip abduction and knee varus moments during stair negotiation are further amplified with asymmetrical loads.

Increases in external knee varus moments have been associated with development of chronic knee pain and asymptomatic medial knee osteoarthritis (Amin et al., 2004). Higher knee varus moments may increase compression of the medial knee joint compartment during gait and may be associated with knee osteoarthritis development (Baliunas et al., 2002). For example, a five year follow-up study indicated that higher knee varus moments resulted in thinning cartilage of the knee joint (Chehab et al., 2014). In addition, higher hip abduction moments may be related to changes in cartilage and greater incidence of hip osteoarthritis (Royer & Wasilewski, 2006). As the increases in knee varus and hip abduction moments were observed during only two steps, it remains to be seen if repetitive cycles of carrying a 20% BW unilateral load may result in higher risk of knee and hip joint injury and cumulative cartilage damage.

When carrying unilateral loads, peak hip abduction and external knee varus moments occurred during step one of stair ascent and descent (Figures 3, 4). Participants contacted step one with their left leg, which is the side opposite of where the unilateral load was carried. This finding may be explained by a larger frontal moment arm from the load in the right hand to the center of pressure under the left foot (MacKinnon & Winter, 1993). Similarly, DeVita et al. (1991) found higher hip abduction and knee varus moments in the leg opposite the load during walking with a sidepack. Thus, it appears that the knee and hip are exposed to the highest frontal plane moments in the leg opposite the load during unilateral load carriage on stairs. As
previously mentioned, additional tests alternating lead legs are needed to account for possible differences between step one and two.

There are several limitations of the current study. One of the limitations is that only two steps of stair negotiation were examined and double limb support was not included in the L5/S1 lateral bending moment. Another limitation is the potentially factors of step (step1 vs. step2) and limb (unloaded limb vs. loaded limb) that may influence the frontal joint moments. For instance, the initial step of stair negotiation can be considered a ‘transition step’ where mechanical demand increases (Christina & Cavanagh, 2002). Also, when carrying an asymmetric load, the frontal joint moments may be sensitive to the individual limb stance because of different moment arms from the center of pressure to the joint centers. Therefore, a further study should focus on these effects by testing both lead legs for each condition.

In summary, there were significant differences in low back and lower extremity moments when comparing load conditions. The 20% BW unilateral load resulted in higher L5/S1 lateral bending, hip abduction, and external knee varus moments than nearly all other loading conditions during stair ascent and descent. Therefore, we suggest potential benefits when carrying symmetric loads in order to decrease the frontal plane joint moments, particularly at the level of 20% BW loads.
References


Table 1. Peak joint moments (mean ± standard deviation) as a function of loading condition during stair ascent and stair descent.

<table>
<thead>
<tr>
<th>Joint Moment (N/kg)</th>
<th>Stair Ascent</th>
<th>Stair Descent</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No load</td>
<td>10% BW bilateral</td>
</tr>
<tr>
<td>L5/S1 Lateral Bending</td>
<td></td>
<td></td>
</tr>
<tr>
<td>No load</td>
<td>0.43±0.11</td>
<td>0.46±0.11</td>
</tr>
<tr>
<td>10% BW bilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>10% BW unilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>20% BW bilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>20% BW unilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip Abduction</td>
<td></td>
<td></td>
</tr>
<tr>
<td>No load</td>
<td>0.98±0.14</td>
<td>1.07±0.19&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>10% BW bilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>10% BW unilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>20% BW bilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>20% BW unilateral</td>
<td></td>
<td></td>
</tr>
<tr>
<td>External Knee Varus</td>
<td></td>
<td></td>
</tr>
<tr>
<td>No load</td>
<td>0.57±0.18</td>
<td>0.62±0.19&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>10% BW bilateral</td>
<td></td>
<td></td>
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<tr>
<td>10% BW unilateral</td>
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<td>20% BW bilateral</td>
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<td></td>
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<tr>
<td>20% BW unilateral</td>
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</tbody>
</table>

<sup>a</sup> p < 0.05 vs. no load,  <sup>b</sup> p < 0.05 vs. 10% BW bilateral,  <sup>c</sup> p < 0.05 vs. 10% BW unilateral,  <sup>d</sup> p < 0.05 vs. 20% bilateral
Figure 1. Illustration of the five load conditions. No load (left), 10% body weight (BW) bilateral load (center), 20% BW bilateral load (center), 10% BW unilateral load (right), and 20% BW unilateral load (right).
Figure 2. Ensemble curves for L5/S1 lateral bending moments during step one and step two of stair ascent and stair descent. Positive values indicate left bending moments, and negative values indicate right bending moments. Unilateral loads were carried in the right hand.
Figure 3. Ensemble curves for hip abduction moments during step one and step two of stair ascent and stair descent. Positive values indicate hip abduction moments, and negative values indicate hip adduction moments. Unilateral loads were carried in the right hand.
Figure 4. Ensemble curves for external knee varus moments during step one and step two of stair ascent and stair descent. Positive values indicate external knee varus moments, and negative values indicate external knee valgus moments. Unilateral loads were carried in the right hand.
CHAPTER 3: CARRYING ASYMMETRICAL LOADS DURING STAIR NEGOTIATION: LOADED LIMB STANCE VS. UNLOADED LIMB STANCE

Abstract

Individuals often carry items in one hand instead of both hands during activities of daily living. The purpose of this study was to investigate low back and lower extremity frontal plane moments for loaded limb stance and unloaded limb stance when carrying symmetric and asymmetric loads during stair negotiation. Participants were instructed to ascend and descend a three-step staircase at preferred pace using a right leg lead and a left leg lead for each load condition: no load, 20% body weight (BW) bilateral load, and 20% BW unilateral load. L5/S1 contralateral bending, hip abduction, external knee varus, and ankle inversion moments were calculated using inverse dynamics. Peak L5/S1 contralateral bending moments were significantly higher when carrying a 20% BW unilateral load as compared to a 20% BW bilateral load for both stair ascent and stair descent. In addition, peak L5/S1 contralateral bending moments were significantly higher during step one than for step two. Peak external knee varus and hip abduction moments were significantly higher in unloaded limb stance as compared to loaded limb stance, specifically when carrying a 20% BW unilateral load. General load carriage recommendations include carrying less than 20% BW loads and splitting loads bilaterally when feasible.
Introduction

Individuals are frequently required to carry heavy items (e.g., suitcases or grocery bags) in one hand during activities of daily living. Carrying heavy loads in one hand can result in adverse changes in posture and how loads are distributed throughout the body during locomotion. Previous studies have reported that asymmetric load carriage increased trunk lateral bending angles (Pascoe et al., 1997; Zhang et al., 2010) and levels of perceived low back pain (Macias et al., 2008). Other studies have found that carrying asymmetric loads with a bag resulted in higher hip abduction moments (Matsuo et al., 2008), a sidepack resulted in higher hip abduction and L5/S1 contralateral bending moments (DeVita et al., 1991), and a hockey bag resulted in increased lower extremity muscle activation (Corrigan & Li, 2014). Therefore, asymmetric load carriage appears to increase frontal plane loading in both the low back and lower extremity.

When carrying a hand-held load on one side of the body, there is a larger moment arm from the load to the stance leg on the opposite side of the body (unloaded limb stance) as compared to the moment arm from the load to the stance leg on the same side of the body (loaded limb stance). Therefore, it is of interest to investigate if frontal plane joint moments are increased during unloaded limb stance. Matsuo et al. (2008) found higher hip abduction moments in unloaded limb stance when carrying a bag, and DeVita et al. (1991) found higher hip abduction and external knee varus moments during unloaded limb stance when carrying a sidepack during walking. These studies investigated asymmetrical load carriage during walking, while the effects of asymmetrical load carriage on unloaded and loaded limb stance during stair negotiation remain unknown.
Stair ascent and descent require higher knee range of motion and knee extension moments than walking (Jevsevar et al., 1993; Riener et al., 2002). Fewer studies have investigated load carriage during stair negotiation. For example, Hong and Li (2005) found that vertical ground reaction forces were significantly higher for a 15% body weight (BW) load during stair descent and a 10% BW load during stair ascent when carrying asymmetric loads in a one-strap athletic bag. Hall et al. (2013) reported significantly higher external knee varus moments when carrying symmetric loads of approximately 20% BW during stair ascent as compared to walking and stair descent. These findings support the ideas that stair negotiation is more demanding on the knees than walking, load carriage increases overall loading on the body, and asymmetric load carriage may further increase frontal plane knee joint moments.

The purpose of this study was to investigate low back and lower extremity frontal plane moments for loaded limb stance and unloaded limb stance when carrying symmetric and asymmetric loads during stair negotiation. We hypothesized that 1) peak external knee varus, hip abduction, and L5/S1 contralateral bending moments would be increased during unilateral load carriage as compared to bilateral load carriage, and 2) peak external knee varus and hip abduction moments would be significantly higher during unloaded limb stance as compared to loaded limb stance during unilateral load carriage.

Methods

Twenty-three healthy young adults with an age range of 20 to 30 (11 males and 12 females; age 21.8 ± 2.4 years; height 173.3 ± 8.8 cm; mass 72.6 ± 12.6 kg) participated in this study. Participants were free of any pathology that would affect them while walking on stairs or prevent them from being able to carry a 20% BW load. Individuals were excluded if they had
back, neck, leg, foot, or arm pain. Prior to participating in the study, each subject read and signed an informed consent document approved by the university’s institutional review board.

Three load conditions were tested: no load, 20% BW bilateral load, and 20% BW unilateral load (Figure 1). The load was evenly split between the right and left hands during the bilateral load condition (10% BW in each hand). Two hand-held bags were utilized in this study and filled with sealed bags of lead shot to match the two loaded conditions. The unilateral load was carried in the participant’s dominant hand. Since all participants were right-handed, they carried the hand-held bag in the right hand during the unilateral load condition. The load carried in the bags was normalized according to each participant’s body weight. The level of normalized load was based on the upper range of previous studies that indicated significant kinematic and/or kinetic changes when carrying loads ranging from 10% to 20% BW (DeVita et al., 1991; Chow et al., 2005; Hong & Li, 2005; Fowler et al., 2006).

Participants were instructed to ascend and descend a three-step staircase (step height 18.5 cm, tread depth 29.5 cm) at preferred pace using a right leg lead and a left leg lead for each load condition (no load, 20% BW bilateral load, 20% BW unilateral load). The order of the conditions was randomized, and each condition was repeated three times for a total of 36 trials (3 load conditions x ascent vs. descent x right vs. left leg leads x 3 trials). Both a right leg and a left leg lead were tested to avoid results being biased by any differences in joint moments that might occur when comparing step one versus step two of stair negotiation.

Eight cameras (Vicon, Los Angeles, CA) were used to collect three-dimensional kinematic data during each testing condition. The dynamic marker set included bilateral great toe, lateral midfoot, lateral malleolus, anterior calf, lateral calf, lateral knee joint line, anterior thigh, lateral thigh, greater trochanter, anterior superior iliac spine (ASIS), posterior superior
iliac spine (PSIS), and acromion process markers, along with a single sacrum and cervical maker. Six additional markers (bilateral heels, medial malleoli, and medial knee joint lines) were recreated using transformations determined from a static standing trial. Two force platforms (AMTI, Watertown, MA) placed on the first and second steps were used to collect kinetic data. Video data were sampled at 160 Hz, and force platform data were sampled at 1600 Hz. Video and force platform data were synchronized using Vicon Nexus (Vicon, Los Angeles, CA).

Data processing

Video and force platform data were processed with a fourth-order, symmetric low-pass Butterworth filter at a cut-off frequency of 6 Hz. The force data were downsampled from 1600 Hz to 160 Hz. Segment masses, center of mass locations, and moments of inertia were scaled to participant anthropometrics (De Leva, 1996). Frontal plane moments were of interest for comparing symmetric and asymmetric loads. L5/S1 contralateral/ipsilateral bending, hip abduction/adduction, external knee varus/valgus, and ankle eversion/inversion moments were calculated using inverse dynamics and rigid body assumptions. The location of the L5/S1 joint center was defined as 34% of the distance from the sacrum marker to the midpoint of the ASIS markers (de Looze et al., 1992; Gillette et al., 2009).

L5/S1 lateral bending moments were analyzed during the single stance phase of the first and second steps since both legs were not always positioned on a force platform. A positive L5/S1 lateral bending moment was toward the left side of the body, which corresponds to the unloaded stance leg or contralateral side of the body during unilateral load carriage. Hip abduction and external knee varus moments were analyzed during the entire stance phase of the first and second steps. Peak joint moments were determined and normalized by body mass. All calculations were performed using a custom Matlab code.
After checking assumptions of multivariate normality, correlations, and sphericity, the effect of the three load conditions and the effect of step one vs. step two on peak L5/S1 contralateral bending, hip abduction, external knee varus, and ankle inversion moments were analyzed using a repeated measures Multivariate Analysis of Variance (3×2 MANOVA). Univariate repeated measures Analyses of Variance (ANOVAs) were performed when main effects of the MANOVA were significant. Bonferroni post-hoc adjustments were used to adjust for multiple comparisons. To investigate the effect of the loaded limb vs. the unloaded limb stance during 20% unilateral load carriage, the Hotelling test was performed. Paired t-tests were used when a main effect of the Hotelling test was significant. The level of statistical significance for all tests was set at $\alpha < 0.05$. Statistical analyses were performed using SPSS (version 23; SPSS Inc., Chicago, IL, USA).

Results

Stair ascent

The MANOVA revealed a significant main effect of load condition ($p < 0.001$). Univariate ANOVA indicated main effects of load condition on peak L5/S1 contralateral bending, hip abduction, external knee varus, and ankle inversion moments (Table 1). L5/S1 contralateral bending, hip abduction, and external knee varus moments were higher for the 20% unilateral load than the 20% bilateral load or no load ($p \leq 0.016$). In addition, hip abduction, external knee varus, and ankle eversion moments were higher for the 20% bilateral load than no load ($p \leq 0.002$). The MANOVA also revealed a main effect of step number ($p < 0.001$). Univariate ANOVA indicated that L5/S1 contralateral bending, hip abduction, external knee varus, and ankle inversion moments were higher during step 1 than step 2 (Table 1).
In addition, a significant interaction of load condition and step number was found for peak L5/S1 contralateral bending moments (Table 1). Therefore, simple effects for each combination of load condition and step number were tested. L5/S1 contralateral bending moments were higher for the 20% unilateral load than the 20% bilateral load and no load for both step 1 and step 2 ($p < 0.001$, Figure 2a). L5/S1 contralateral bending moments were also significantly higher for step 1 than step 2 for all load conditions: no load, 20% BW bilateral load, and 20% unilateral load ($p < 0.001$). Therefore, the main effects held true for all interaction combinations.

The Hotelling test revealed a significant main effect of stance limb ($p < 0.001$). Paired t-tests indicated that external knee varus and hip abduction moments were higher for the unloaded stance limb than the loaded stance limb ($p < 0.001$, Figure 4a). Ensemble curves illustrating external knee varus and hip abduction moments of the unloaded and loaded stance limb during stair ascent are shown in Figure 5.

**Stair descent**

The MANOVA revealed a significant main effect of load condition ($p < 0.001$). Univariate ANOVA indicated main effects of load condition on L5/S1 contralateral bending, hip abduction, external knee varus, and ankle inversion moments (Table 2). L5/S1 contralateral bending, hip abduction, external knee varus, and ankle inversion moments were higher for the 20% unilateral load than no load ($p < 0.001$). L5/S1 contralateral bending moments were also higher for the 20% unilateral than the 20% bilateral load ($p < 0.001$). In addition, hip abduction and external knee varus moments were higher for the 20% bilateral load than no load ($p < 0.001$). The MANOVA also revealed a main effect of step number ($p < 0.001$). Univariate ANOVA indicated that L5/S1 contralateral bending moments were higher during step 1 than step
2, while hip abduction and ankle inversion moments were higher during step 2 than step 1 (Table 2).

In addition, a significant interaction of load condition and step number was found for peak L5/S1 contralateral bending and hip abduction moments (Table 2). L5/S1 contralateral bending moments were higher for the 20% unilateral load than the 20% bilateral load and no load for both step 1 and step 2 ($p < 0.001$, Figure 2b). L5/S1 contralateral bending moments were also higher for step 1 than step 2 for all load conditions: no load, 20% BW bilateral load, and 20% unilateral load ($p < 0.001$). Therefore, L5/S1 contralateral bending moment main effects held true for all interaction combinations. Hip abduction moments were only higher for step 2 than step 1 for the 20% BW unilateral load ($p < 0.001$), but not for the 20% bilateral load or no load (Figure 3).

The Hotelling test revealed a significant main effect of stance limb ($p < 0.001$). Paired t-tests indicated that peak L5/S1 contralateral bending moments were higher for the loaded stance limb, while external knee varus and hip abduction moments were higher for the unloaded stance limb ($p < 0.001$, Figure 4b). Ensemble curves illustrating L5/S1 contralateral bending, external knee varus, and hip abduction moments for the unloaded and loaded stance limb are shown in Figure 6.

Discussion

The purpose of this study was to investigate the effect of load symmetry on loaded limb and unloaded limb stance low back and lower extremity frontal plane moments for loaded limb stance and unloaded limb stance when carrying symmetric and asymmetric loads during stair ascent and stair descent.
Stair ascent

Our first hypothesis that peak external knee varus, hip abduction, and L5/S1 contralateral bending moments would be increased during unilateral load carriage as compared to bilateral load carriage was supported during stair ascent (Table 1). As expected, these results indicate that the external load imbalance introduced by unilateral load carriage is reflected in the frontal plane joint moments. The L5/S1 appeared to be particularly sensitive to load asymmetry as the peak contralateral moments were unchanged when comparing a 20% BW bilateral load to no load. Increased L5/S1 lateral bending moments may lead to low back pain or injury. For example, increases in lateral bending moments are associated with increased compressive and shear forces on the intervertebral discs and ligaments (Marras & Granata, 1997). McGill et al. (2013) reported that asymmetric load carriage with a 30 kg bucket in one hand resulted in higher compressive spinal loading (2800 N) as compared to bilateral load carriage with 30 kg buckets in both hands (1570 N) during walking. Schmidt et al. (2007) demonstrated that lateral bending moments combined with axial moments may contribute to failure of intervertebral discs. Therefore, our results suggest that when feasible, it is beneficial to split a unilateral load into bilateral loads to reduce low back loading.

Our second hypothesis that peak external knee varus and hip abduction moments would be higher for unloaded limb stance during unilateral load carriage was supported for stair ascent (Figure 4a). In fact, peak hip abduction moments were over 100% higher and peak external knee varus moments were over 200% higher during unloaded limb stance. These results are likely explained by the greater frontal plane moment arms from the center of pressure of the unloaded limb to the unilaterally carried load. Similar patterns have also been observed when carrying a 20% BW one-strap sidepack during normal walking (DeVita et al., 1991). Also, Neumann
(1996) reported higher hip abductor muscle activation in the unloaded limb when carrying 15% BW asymmetric loads in one hand as compared to the unloaded condition. This effect appeared to be limited to the lower extremity as L5/S1 bending moments were not significantly higher during unloaded leg stance. Increased knee external varus moments are associated with increased compressive loading in the medial knee joint compartment (Baliunas et al., 2002), and thus can be of particular concern for development of chronic knee pain or osteoarthritis (Amin et al., 2004). A finite element study demonstrated that increased knee external varus moments also resulted in higher ACL strain (Bendjaballah et al., 1997). In addition, increased hip abduction moments may result in higher compressive forces in the articular joint surface and lead to joint degeneration (Neumann, 1989). Therefore, the results suggest that asymmetric load carriage tasks may involve higher injury risk for the unloaded leg during stair ascent.

Peak frontal plane moments were higher during the first step than the second step of stair ascent (Table 1, Figure 2). The initial step of stair ascent is often considered a transition from standing or walking to a repeating pattern of stair negotiation. Peak joint moments during the first step suggest the importance of measuring this transition when analyzing stair ascent. Taken together, L5/S1 contralateral bending moments were highest when carrying a 20% BW unilateral load during step one. Ensemble curves further illustrate that the peak external knee varus and hip abduction moments occurred in the unloaded leg when carrying a 20% BW unilateral load during 20-40% of stance during step one (Figure 5).

Stair descent

Our first hypothesis that peak frontal plane moments would be increased during unilateral load carriage as compared to bilateral load carriage was only supported for L5/S1 contralateral bending moments during stair descent (Table 2). Similar to stair ascent, the L5/S1 appeared to be
particularly sensitive to load asymmetry as the peak contralateral moments were unchanged when comparing a 20% BW bilateral load to no load. When examining Table 1 and Table 2, stair descent resulted in higher peak frontal plane moments than stair ascent for all loading conditions. These results suggest the relative difficulty of both loaded and unloaded stair descent as compared to stair ascent.

Our second hypothesis that peak external knee varus and hip abduction moments would be higher for unloaded limb stance during unilateral load carriage was also supported for stair descent (Figure 4b). The differences were once again substantial, with peak hip abduction moments 140% higher and peak external knee varus moments over 200% higher during unloaded limb stance. As with stair ascent, these results are likely explained by the larger frontal plane moment arms from the center of pressure of the unloaded limb to the unilateral load. Thus, unilateral load carriage would produce asymmetric joint loading between the loaded and unloaded limbs, which may lead to pathologic changes in lower extremity joints and higher incidence of knee and hip osteoarthritis (Neumann, 1989; Amin et al., 2004).

In contrast, peak L5/S1 contralateral bending moments were significantly higher during the loaded limb stance. This appears to indicate that the lower extremity and low back play different roles in adjusting to asymmetric loads during stair descent, with the lower extremity playing a larger role during unloaded limb stance and the low back playing a larger role during loaded limb stance. For instance, hip abduction moments may be utilized to maintain the center of mass (COM) within the base of support (MacKinnon & Winter, 1993) during the unloaded stance. On the other hand, upper body adjustments may be required when the COM is close to the base of support during the loaded limb stance when carrying asymmetric loads.
Peak L5/S1 contralateral bending moments were higher during the first step, while peak hip abduction moments were higher during the second step of stair descent (Table 2, Figure 2). These results may further indicate the different roles of the lower extremity and low back, but the effect of unloaded versus loaded limb stance was much greater than the effect of step number for hip abduction moments. Ensemble curves illustrate that L5/S1 contralateral bending moments were highest in loaded limb stance when carrying a 20% BW unilateral load during step one of stair descent (Figure 6). Combining factors, peak hip abduction moments were highest in unloaded limb stance when carrying a 20% BW load (unilateral or bilateral) during step two.

There are several limitations of the current study. One of the limitations is that a three step staircase was used, so the participants may not have achieved a repeatable stair negotiation pattern. However, the results indicated the importance of considering the first transition step of stair ascent and descent. If considering load carriage guidelines, another limitation is that only no load and 20% BW loads were tested. With significant differences occurring at with a 20% BW loads, it is unclear if a 10% BW or 15% BW load would have also resulted in significant differences.

In summary, there are several primary concerns when considering load carriage during both stair ascent and stair descent:

- L5/S1 contralateral bending moments when carrying a 20% BW unilateral load
- External knee varus and hip abduction moments in unloaded limb stance, specifically when carrying a 20% BW unilateral load
- L5/S1 contralateral bending moments during step one
General load carriage recommendations include carrying less than 20% BW loads and splitting loads bilaterally. Assessment recommendations include also analyzing the first step of negotiation and analyzing both the loaded and unloaded limbs.
References


Figure 1. Illustration of the three load conditions: no load (left), 20% BW bilateral load split between both sides of the body (center), and 20% body weight (BW) load on one side of the body (right).
Table 1. Peak mean and S.D joint moments during stair ascent and statistical results from univariate ANOVAs (Load × Step)

<table>
<thead>
<tr>
<th></th>
<th>Stair ascent</th>
<th>Main effect</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No load Mean(SD)</td>
<td>Bilateral Mean(SD)</td>
</tr>
<tr>
<td>L5S1 contralateral bending (Nm/kg)</td>
<td>0.160 (.109)</td>
<td>0.153 (.118)</td>
</tr>
<tr>
<td>Hip abduction (Nm/kg)</td>
<td>0.817 (.121)</td>
<td>0.967a (.143)</td>
</tr>
<tr>
<td>External knee varus (Nm/kg)</td>
<td>0.455 (.129)</td>
<td>0.521a (.142)</td>
</tr>
<tr>
<td>Ankle inversion (Nm/kg)</td>
<td>0.104 (.048)</td>
<td>0.134a (.071)</td>
</tr>
<tr>
<td>L5S1 ipsilateral bending (Nm/kg)</td>
<td>0.326 (.089)</td>
<td>0.354 (.107)</td>
</tr>
<tr>
<td>Hip adduction (Nm/kg)</td>
<td>0.103 (.032)</td>
<td>0.139 (.041)</td>
</tr>
<tr>
<td>External knee valgus (Nm/kg)</td>
<td>0.076 (.031)</td>
<td>0.096 (.041)</td>
</tr>
<tr>
<td>Ankle eversion (Nm/kg)</td>
<td>0.053 (.029)</td>
<td>0.061 (.029)</td>
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</tbody>
</table>

*indicates p < 0.05; ‘a’ indicates a significant difference when compared to no load; ‘b’ indicate a significant difference when compared to 20% BW bilateral loads.
<table>
<thead>
<tr>
<th></th>
<th>Stair descent</th>
<th>Main effect</th>
<th>LOAD × STEP</th>
<th>LOAD</th>
<th>STEP</th>
<th>LOAD × STEP</th>
</tr>
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<tbody>
<tr>
<td></td>
<td>No load Mean(SD)</td>
<td>Bilateral Mean(SD)</td>
<td>Unilateral Mean(SD)</td>
<td>Step1 Mean(SD)</td>
<td>Step2 Mean(SD)</td>
<td>p (power)</td>
</tr>
<tr>
<td><strong>L5S1 contralateral bending</strong></td>
<td><strong>(Nm/kg)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip abduction (Nm/kg)</td>
<td>1.013 (.105)</td>
<td>1.192a (.113)</td>
<td>1.166a (.144)</td>
<td>1.094 (.141)</td>
<td><strong>1.153</strong> (.143)</td>
<td>&lt; <strong>0.001</strong></td>
</tr>
<tr>
<td>External knee varus (Nm/kg)</td>
<td>0.591 (.141)</td>
<td>0.689a (.145)</td>
<td>0.676a (.189)</td>
<td>0.669 (.171)</td>
<td>0.635 (.157)</td>
<td>&lt; <strong>0.001</strong></td>
</tr>
<tr>
<td>Ankle inversion (Nm/kg)</td>
<td>0.127 (.054)</td>
<td>0.139 (.051)</td>
<td>0.145a (.054)</td>
<td>0.127 (.054)</td>
<td><strong>0.146</strong> (.051)</td>
<td><strong>0.003</strong></td>
</tr>
<tr>
<td><strong>L5S1 ipsilateral bending</strong></td>
<td><strong>(Nm/kg)</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip adduction (Nm/kg)</td>
<td>0.087 (.046)</td>
<td>0.089 (.044)</td>
<td>0.087 (.039)</td>
<td>0.066 (.028)</td>
<td>0.108 (.045)</td>
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<tr>
<td>External knee valgus (Nm/kg)</td>
<td>0.046 (.039)</td>
<td>0.040 (.030)</td>
<td>0.092 (.064)</td>
<td>0.051 (.042)</td>
<td>0.067 (.059)</td>
<td>-</td>
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<tr>
<td>Ankle eversion (Nm/kg)</td>
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<td>0.131 (.053)</td>
<td>0.138 (.054)</td>
<td>0.126 (.045)</td>
<td>0.130 (.056)</td>
<td>-</td>
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</table>

*indicates p < 0.05; ‘a’ indicates a significant difference when compared to no load; ‘b’ indicate a significant difference when compared to 20% BW bilateral loads.
Figure 2. Results of simple main effects for L5/S1 contralateral bending moment during stair ascent (A) and descent (B). The simple main effects were tested for each load condition and each step number due to a significant interaction. * indicates a significant difference between step 1 and step 2; ‘a’ indicates a significant difference when compared to no load; ‘b’ indicates a significant difference when compared to 20% BW bilateral load.
Figure 3. Results of simple main effects for hip abduction moment during stair descent. The simple main effects were tested for each load condition and each step number due to a significant interaction. * indicates a significant difference between step 1 and step 2.
Figure 4. Effects of loaded limb stance vs. unloaded limb stance on external knee varus, hip abduction, and L5/S1 contralateral bending moments during ascent (A) and descent (B). * indicates a significant difference (p < 0.001)
Figure 5. Ensemble curves of external knee varus, hip abduction, and L5/S1 contralateral bending moments for 20% BW unilateral load during stair ascent. No significant effects of loaded limb vs. unloaded limb were found for L5S1 bending moment.
Descent

Figure 6. Ensemble curves of external knee varus, hip abduction, and L5/S1 contralateral bending moments for 20% BW unilateral load during stair descent.
CHAPTER 4: CARRYING ASYMMETRIC LOADS WHILE WALKING ON AN UNEVEN SURFACE

Abstract

Asymmetric load carriage is expected to produce a lateral shift of the center of mass and may result in a challenge to postural control. The purpose of this study was to assess postural stability in the medial-lateral (ML) direction when carrying unilateral versus bilateral loads and when walking on even versus uneven surfaces. Nineteen healthy young adults walked on even and uneven surface treadmills under three load conditions: no load, 20% body weight (BW) bilateral load, and 20% BW unilateral load. There dependent variables were: double stance ratio, ML center of pressure (COP) excursion, ML COP velocity, ML time-to-contact (TTC) percentage, coefficient of variation (CV) of double stance ratio, CV of ML COP excursion, and CV of ML COP velocity. Carrying 20% BW bilateral or unilateral loads resulted in a significantly higher double stance ratio, consistent with a more conservative postural strategy. Unilateral load carriage further resulted in a significantly increased CV of double stance ratio, indicating higher variability within that more conservative postural strategy. Walking on an uneven surface also resulted in a significantly increased CV of double stance ratio. Unloaded limb stance showed a higher double stance ratio and ML COP velocity when carrying a 20% BW unilateral load, although it appears that the loaded limb may be used to make step-by-step adjustments as evidenced by the higher CV of ML COP velocity. Therefore, unilateral load carriage, walking on uneven surfaces, and balance on the unloaded leg are of particular concern when considering postural stability.
Introduction

Individuals often carry items in one hand instead of both hands during activities of daily living. Asymmetric load carriage is expected to produce a lateral shift of the center of mass and may result in a challenge to postural control during walking (DeVita et al., 1991). Previous studies have reported that an asymmetric load with a sidepack or a briefcase resulted in higher medial-lateral (ML) center of pressure (COP) velocities during quiet standing (Zultowski & Aruin, 2008) and greater ML COP displacement during gait initiation (Vieira et al., 2016). However, a recent study reported no differences between bilateral and unilateral shopping bags in ML COP excursion and velocity during quite standing (Bampouras & Dewhurst, 2016). The discrepancy in results between previous studies may be due to static versus dynamic testing conditions and indicates that further study is need to determine how unilateral load carriage affects postural stability during walking.

Individuals may carry asymmetric loads over a challenging surfaces such as uneven or irregular terrain, which may require a higher demand for postural control than walking on an even surface. For example, Thies et al. (2005) found increased step width and step time variability while walking on an irregular surface, Gates et al. (2013) reported greater step width variability and higher ML center of mass (COM) velocity while walking on a loose rock surface, and Marigold and Patla (2008) found greater step and ML trunk bending variability while walking on a multi-surface terrain. Previous studies have also demonstrated that asymmetric load carriage resulted in significant differences in lower extremity joint moments between loaded and unload limbs (DeVita et al., 1991; Matsuo et al., 2008). These findings support the idea that the uneven surfaces are more challenging for postural control, particularly in the ML direction, and
that it is of interest to investigate differences for postural control between the loaded versus unloaded limb stance.

Postural stability can be evaluated by how individuals control the COP within the base of support of the foot. Time-to-contact (TTC) is the estimated time it takes the COP to reach the boundary of foot (Slobounov et al., 1998). TTC includes both spatial and temporal (COP position, velocity and acceleration) aspects of postural control relative to the base of support (Haddad et al., 2006). Time-to-contact (TTC) has been used to evaluate postural stability and provide a measure of how long the individual has to make a postural adjustment before reaching postural instability (Slobounov et al., 1998; Haddad et al., 2006; Gruber et al., 2011). However, this approach has been limited to static tasks such as quiet standing. A standard TTC analysis is challenging to apply to gait since the COP must leave the boundary of one foot and shift to the other foot as human body progresses forward. Therefore, a modified TTC method was proposed to evaluate postural stability during walking in the current study.

It is believed that increasing double stance time is a common strategy to improve postural stability during challenging walking conditions or unstable gait. Several studies have reported that symmetric load carriage with a backpack resulted in increased double stance time during walking (Wang et al., 2001; Chow et al., 2005). In addition, increased double support time may be utilized to improve postural stability for individuals at high risk for falls (Maki, 1997; Gok et al., 2002). Thus, increased double support time may be indicative of an attempt to improve postural stability and to avoid loss of balance during challenging walking conditions.

Many studies have investigated how load carriage affects the biomechanics of human movement as a function of load magnitude. However, we are not aware of any studies that have evaluated postural stability of unilateral load carriage while walking on an uneven surface.
Therefore, the purpose of this study was to assess postural stability, particularly in the ML direction, when carrying unilateral versus bilateral loads and when walking on even versus uneven surfaces. Previous studies have indicated that asymmetric load carriage (Zultowski & Aruin, 2008; Vieira et al., 2016), walking on uneven surfaces (Thies et al., 2005; Marigold & Patla, 2008; Gates et al., 2013), and unloaded limb stance (DeVita et al., 1991) present postural challenges and/or loading asymmetry in the ML direction. We hypothesized that ML COP velocity would be increased and ML TTC percentage would be decreased 1) during unilateral load carriage as compared to bilateral load carriage and 2) when walking on an uneven surface as compared to an even surface, and 3) during unloaded limb stance as compared to loaded limb stance for unilateral load carriage.

Methods

Nineteen healthy young adults with an age range of 18 to 30 (14 males and 5 females; age 25.5 ± 3.9 years; height 172.6 ± 5.0 cm; mass 69.7 ± 7.2 kg) participated in this study. Participants were free of any pathology that would affect them while walking on a treadmill or prevent them from being able to carry a 20% body weight (BW) load. Individuals were excluded if they had back, neck, leg, foot, or arm pain. Prior to participating in the study, each participant read and signed an informed consent form approved by the university’s institutional review board.

Three load conditions were tested: no load, 20% BW bilateral load, and 20% BW unilateral load (Figure 1). The 20% unilateral load was carried in the participant’s dominant hand, while the 20% BW bilateral load was evenly split between both sides of the body (10% BW load carried with each hand). Two hand-held bags were utilized in this study and filled with sealed bags of lead shot to match a load normalized according to each participant’s body weight.
These normalized loads were based on previous studies that indicated significant kinematic and/or kinetic changes when carrying loads ranging from 10% to 20% BW (DeVita et al., 1991; Hong et al., 2000; Chow et al., 2005; Hong & Li, 2005; Fowler et al., 2006). Two different treadmills were used for the even and uneven surface conditions (Figure 2). The participants completed six total conditions (2 surfaces ×3 loads).

Initially, the participants were instructed to walk on both even and uneven surface treadmills for one minute under the three load conditions as a warm-up session. This session allowed the subjects to familiarize with the load carriage conditions (Graham et al., 2015). The treadmill velocity was started at 0.22 m/s and then the speed was gradually increased or decreased until the participant signaled that the preferred walking speed had been reached. The six preferred walking speeds were recorded and then the slowest walking speed was selected as a constant walking speed for further data collection. The participants were then asked to walk on the even and uneven surface treadmills for 90 seconds under the three load conditions in a randomized order. Each participant was allowed to rest as much as necessary between conditions, with a minimum break of one minute. A Pedar in-shoe pressure system (Novel, Munich, Germany) was used to collect vertical forces and COP in each foot at 100 Hz.

Data processing

Ten strides were selected and analyzed during the last thirty seconds of each condition. During unilateral load carriage, the loaded limb was on the side of the carried load, while the unloaded limb was the opposite side. Single stance phases and double stance phases were determined using the vertical forces, and double stance ratio was calculated as a ratio of double stance time to single stance time. A rectangular base of support for each foot (Figure 3) was defined by the dimensions of the Pedar insole sensor (85 mm × 260 mm or 270 mm). The origin
of the insole sensor is located at the most posterior and medial point of the sensor, and thus the COP positions were recoded as anterior-posterior and ML coordinates relative to this sensor origin. ML COP excursion and mean ML COP velocity were determined during single stance phases for each foot. ML COP velocities and accelerations were calculated utilizing the first central difference method.

ML COP positions, velocities, and accelerations were used to calculate ML TTC using the equation in Figure 3. Since the COP shifts between the boundaries of each foot during walking, the assessment of TTC commonly used during quiet stance was modified. TTC was calculated at each data point and then compared to the remaining single stance time (Figure 4). If the TTC was less than the remaining single stance time, then the TTC value was stored for that time point, indicating a postural adjustment was required during single stance. If the TTC was greater than the remaining single stance time, then the TTC was set to the remaining single stance time, indicating that the other foot would contact with the ground in the dual support phase before a postural adjustment was needed. TTC percentage was then calculated by normalizing TTC by mean remaining single stance time during each single stance. A TTC percentage of 100% indicated that no postural adjustment was required during single stance.

Variability in double stance ratio, ML COP excursion, and ML COP velocity was evaluated though coefficient of variation (CV) for ten strides. Increased stride-to-stride variability can be indicative of inconsistent steps and decreased stability during walking (Hausdorff et al., 1997). In total, there were seven dependent variables: double stance ratio, ML COP excursion, mean ML COP velocity, ML TTC percentage, CV of double stance ratio, CV of ML COP excursion, and CV of ML COP velocity. COP-based parameters were calculated using a custom-made Matlab code (Mathworks Inc., Natik, MA).
Statistical analyses

The effects of the different loading conditions and the effects of different surfaces on COP parameters were analyzed using repeated measures Multivariate Analysis of Variance (3×2 MANOVA). Separate MANOVAs were performed for right and left limbs. Univariate repeated measures Analysis of Variance (ANOVA) was performed when main effects of the MANOVA were significant. Bonferroni post-hoc adjustments were used. To investigate the effect of loaded limb stance vs. unloaded limb stance, the Hotelling test was performed. Paired t-tests were performed when a main effect of the Hotelling test was significant. The level of statistical significance for all tests was set at $\alpha < 0.05$. Statistical analyses were performed using SPSS® statistics (version 23; SPSS Inc., Chicago, IL, USA).

Results

Effect of load condition

MANOVA revealed significant main effects of load condition for right and left limbs ($p < 0.001$). Univariate ANOVA indicated main effects of load condition on double stance ratio, ML COP excursion, and CV of double stance ratio for both right and left limb stance (Table 1). Double stance ratios were significantly higher when comparing the 20% BW bilateral load and the 20% BW unilateral load to no load during both right and left limb stance ($p \leq 0.001$). In addition, double stance ratios were significantly higher when comparing the 20% BW bilateral load to the 20% BW unilateral load during right limb stance ($p = 0.020$). ML COP excursions were significantly higher when comparing no load to the 20% BW bilateral load during both right and left single limb stances ($p \leq 0.048$) as well as to the 20% BW unilateral load during right single limb stance ($p = 0.019$). CV of double stance ratio was significantly higher for the 20% BW unilateral load than no load during both right and left limb stances ($p \leq 0.030$).
Univariate ANOVA also indicated significant main effects on ML TTC percentage during left single limb stance and CV of ML COP velocity during right single limb stance. ML TTC percentages were significantly higher when comparing the 20% BW bilateral load to no load during left single limb stance ($p = 0.030$). In addition, CV of ML velocity was significantly higher for the 20% BW unilateral load than no load during right limb stance ($p = 0.007$).

**Effect of surface**

MANOVA also revealed a significant main effect of surface condition ($p \leq 0.012$). Univariate ANOVA indicated that ML COP excursion, ML COP velocity, and CV of double stance ratio were significantly higher when comparing the uneven surface to the even surface during right limb stance (Table 1). In addition, double stance ratio and CV of double stance ratio were significantly higher for the uneven surface than the even surface during left limb stance. No significant interactions of the load and surface were found.

**Effect of loaded limb vs. unloaded limb**

The Hoteling test revealed a significant main effect of loaded vs. unloaded limb stance ($p = 0.026$). Univariate $t$-tests indicated that double stance ratio and ML COP velocity were significantly higher for the unloaded limb stance as compared to the loaded limb stance (Table 2). However, CV of ML COP velocity was significantly higher for the loaded single limb stance than the unloaded single limb stance.

**Discussion**

The purpose of this study was to evaluate postural stability when carrying unilateral versus bilateral loads, walking on even versus uneven surfaces, and during loaded versus unloaded limb stance.
Effect of unilateral versus bilateral loads

The first hypothesis that ML COP velocity would be increased while ML TTC percentage would be decreased during unilateral as compared to bilateral load carriage was not supported (Table 1). Conversely, Zultowski and Aruin (2008) found that 20% BW asymmetric loads with a single strap bag and a briefcase resulted in increased ML COP excursion and velocity as compared to 20% BW symmetric loads with a backpack during quiet standing. In addition, Vieira et al. (2016) reported that a 12% BW asymmetric backpack load resulted in increased ML COP excursion than a 12% BW symmetric backpack load during gait initiation. One potential reason for the difference in findings is that our symmetric loads were carried in the hands and thus were further from the center of mass than a symmetric backpack. In the current study, participants appeared to adjust to both symmetric and asymmetric loading by constraining their ML COP excursion as compared to unloaded walking.

ML TTC percentages were not changed when comparing the unilateral to bilateral load (Table 1). One potential explanation for this result is that the average walking speed (0.71 m/s) of this study was slower than a normal walking speed (1.19 m/s) on a treadmill (Dal et al., 2010). The slow walking pace may have required relatively slow postural adjustments as evidenced by ML TTC percentages approaching 100% across conditions. In addition, participants adjusted to both unilateral and bilateral loads by increasing their double stance ratio. Similarly, previous studies have reported increased double support time when carrying a 15% BW load as compared to unloaded walking (Wang et al., 2001; Chow et al., 2005). However, Bampouras and Dewhurst (2016) reported that a 3 kg asymmetric load did not affect double stance time as compared to no load and bilateral load. This disagreement in findings could be due to the relatively higher loads in the current study (20% BW versus <5% BW). Furthermore, a greater proportion of double
stance phase has been seen in high-risk individuals for falls, including older adults (Maki, 1997) and patients with knee osteoarthritis (Gok et al., 2002). Greater double stance time may be a strategy to improve postural stability during unstable gait. With increased time spent in double stance, participants were less likely to have to make a postural adjustment during single stance.

An interesting finding is that stride-to-stride variability in double stance ratio and COP ML velocity (right leg) were higher for the 20% BW unilateral load than for no load (Table 1). Previous studies have suggested that higher stride-to-stride variability in gait parameters reflects inconsistent stepping and decreased postural stability during walking (Hausdorff et al., 2001; Hollman et al., 2007). In addition, CV of temporal-spatial gait parameters have been frequently used to estimate gait variability associated with increased risk of falling (Grabiner et al., 2001; Hollman et al., 2007; Kang & Dingwell, 2008; Reelick et al., 2009; Rao et al., 2011). In this sense, CV of COP parameters may be a useful estimator of postural stability during load carriage in terms of predicting fall risk. Specifically, an increase in the CV of double stance ratio may suggest decreased postural stability during unilateral load carriage.

Effect of even versus uneven surfaces

The second hypothesis that ML COP velocity would be increased while ML TTC percentage would be decreased when walking on an uneven surface was partially supported. ML COP velocity was higher for the uneven surface (right foot), but there were no differences in ML TTC percentage (Table 1). Similarly, Gates et al. (2013) reported higher ML COM velocity with a rocky surface than an even surface, but no changes in the lateral margin of stability. Thus, COP and COM velocities may be more sensitive to changes than ML TTC percentage, which may be due to ML trunk sway when walking on the uneven surface (Marigold & Patla, 2008). In addition, increases in double stance ratio (left leg), ML COP excursion (right leg), and CV of
double stance ratio were observed with the uneven surface. These results may correspond to a previous study that found increased step width and step time variability with an uneven surface (Thies et al., 2005).

Effect of loaded versus unloaded limb

The third hypothesis that ML COP velocity would be higher while ML TTC percentage would be lower during unloaded versus loaded limb stance was partially supported. ML COP velocity was higher for unloaded limb stance, but ML TTC percentage was not significantly different (Table 2). Since the carried load is further from the unloaded limb, a higher ML COP velocity may be required to shift body weight to the loaded limb. However, the loaded limb stance showed increased CV of ML COP velocity. As an explanation, Dingwell and Marin (2006) suggested that increased stride-to-stride variability can be exchanged to increase local dynamic stability that accompany a slower walking speed. Beauchet et al. (2009) also suggested that decreased walking speed resulted in higher stride time variability. Thus, slower ML COP velocity for the loaded limb stance may be associated with increased variability in ML COP velocity. In addition, the double stance ratio was increased for the unloaded limb during unilateral load carriage, which is associated with balance challenges (Gabell & Nayak, 1984). Therefore, these combined results seem to imply decreased postural stability during the unloaded limb stance.

There are several limitations to this study. First, COP measures are whole body parameters, so without additional video analysis, we are unable to know where in the body postural adjustment are being made. Second, we used treadmills to maintain a consistent, but slow preferred walking velocity, when individuals may adjust their preferred and step-to-step
velocity in non-lab situations. Third, the TTC methods we used to analyze gait are new, so we don’t have previous values to judge ‘good’ and ‘bad’ values for postural stability.

In summary, carrying 20% BW bilateral or unilateral loads resulted in a higher double stance ratio, consistent with a more conservative postural strategy. Unilateral load carriage further resulted in an increased CV of double stance ratio, indicating higher variability within that more conservative postural strategy. Walking on an uneven surface also resulted in an increased CV of double stance ratio. Unloaded limb stance showed a higher double stance ratio and ML COP velocity, although it appears that the loaded limb may be used to make step-by-step adjustments as evidenced by the higher CV of ML COP velocity. Therefore, unilateral load carriage, walking on uneven surfaces, and balance on the unloaded leg are of particular concern when considering postural stability.
References


Table 1. Mean and standard deviations for postural stability parameters under the three load conditions for the even surface and uneven surface.

<table>
<thead>
<tr>
<th></th>
<th>No Load Mean (SD)</th>
<th>Bilateral Mean (SD)</th>
<th>Unilateral Mean (SD)</th>
<th>Even Mean (SD)</th>
<th>Uneven Mean (SD)</th>
<th>Load p-value</th>
<th>Surface p-value</th>
<th>Load *Surf p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Double stance ratio</strong></td>
<td></td>
<td></td>
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<td></td>
</tr>
<tr>
<td>Right Leg</td>
<td>0.458 (0.036)</td>
<td><strong>0.504</strong>&lt;sup&gt;a&lt;/sup&gt; (0.035)</td>
<td>0.489&lt;sup&gt;ab&lt;/sup&gt; (0.036)</td>
<td>0.480 (0.041)</td>
<td>0.488 (0.040)</td>
<td>&lt;0.001</td>
<td>0.149</td>
<td>0.564</td>
</tr>
<tr>
<td>Left Leg</td>
<td>0.458 (0.033)</td>
<td><strong>0.504</strong>&lt;sup&gt;a&lt;/sup&gt; (0.034)</td>
<td>0.509&lt;sup&gt;a&lt;/sup&gt; (0.044)</td>
<td>0.485 (0.046)</td>
<td><strong>0.499</strong>&lt;sup&gt;+&lt;/sup&gt; (0.041)</td>
<td>&lt;0.001</td>
<td>0.033</td>
<td>0.594</td>
</tr>
<tr>
<td><strong>ML COP excursion(mm)</strong></td>
<td></td>
<td></td>
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<td></td>
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<tr>
<td>Right Leg</td>
<td>11.83 (4.31)</td>
<td><strong>10.35</strong>&lt;sup&gt;a&lt;/sup&gt; (4.22)</td>
<td>10.09&lt;sup&gt;a&lt;/sup&gt; (3.87)</td>
<td>10.22 (4.24)</td>
<td>11.30&lt;sup&gt;+&lt;/sup&gt; (4.07)</td>
<td>0.007</td>
<td>0.008</td>
<td>0.805</td>
</tr>
<tr>
<td>Left Leg</td>
<td>11.95 (3.84)</td>
<td><strong>10.11</strong>&lt;sup&gt;a&lt;/sup&gt; (4.33)</td>
<td>11.01 (4.21)</td>
<td>10.71 (4.22)</td>
<td>11.34 (4.12)</td>
<td>0.009</td>
<td>0.299</td>
<td>0.932</td>
</tr>
<tr>
<td><strong>ML COP velocity(mm/s)</strong></td>
<td></td>
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<tr>
<td>Right Leg</td>
<td>26.6 (9.6)</td>
<td>25.7 (10.4)</td>
<td>25.6 (10.1)</td>
<td>23.8 (9.5)</td>
<td><strong>27.9</strong>&lt;sup&gt;+&lt;/sup&gt; (10.1)</td>
<td>0.589</td>
<td>&lt;0.001</td>
<td>0.817</td>
</tr>
<tr>
<td>Left Leg</td>
<td>26.7 (8.6)</td>
<td>25.7 (11.4)</td>
<td>28.5 (11.5)</td>
<td>25.6 (10.3)</td>
<td>28.4 (10.7)</td>
<td>0.085</td>
<td>0.053</td>
<td>0.579</td>
</tr>
<tr>
<td><strong>ML TTC percentage</strong></td>
<td></td>
<td></td>
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<td></td>
</tr>
<tr>
<td>Right Leg</td>
<td>98.7 (1.8)</td>
<td>99.4 (1.0)</td>
<td>98.9 (1.3)</td>
<td>99.1 (1.4)</td>
<td>98.8 (1.5)</td>
<td>0.103</td>
<td>0.273</td>
<td>0.470</td>
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<tr>
<td>Left Leg</td>
<td>98.5 (2.3)</td>
<td><strong>99.4</strong>&lt;sup&gt;a&lt;/sup&gt; (1.4)</td>
<td>98.9 (1.2)</td>
<td>99.0 (2.0)</td>
<td>98.9 (1.3)</td>
<td><strong>0.047</strong></td>
<td>0.830</td>
<td>0.702</td>
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<td><strong>CV double stance ratio</strong></td>
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<td></td>
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</tr>
<tr>
<td>Right Leg</td>
<td>0.044 (0.016)</td>
<td>0.051 (0.024)</td>
<td><strong>0.054</strong>&lt;sup&gt;a&lt;/sup&gt; (0.019)</td>
<td>0.044 (0.019)</td>
<td><strong>0.055</strong>&lt;sup&gt;+&lt;/sup&gt; (0.020)</td>
<td>0.044</td>
<td>&lt;0.001</td>
<td>0.748</td>
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<tr>
<td>Left Leg</td>
<td>0.045 (0.024)</td>
<td>0.049 (0.028)</td>
<td><strong>0.055</strong>&lt;sup&gt;a&lt;/sup&gt; (0.026)</td>
<td>0.042 (0.023)</td>
<td><strong>0.057</strong>&lt;sup&gt;+&lt;/sup&gt; (0.027)</td>
<td>0.012</td>
<td>&lt;0.001</td>
<td>0.298</td>
</tr>
<tr>
<td><strong>CV ML COP excursion</strong></td>
<td></td>
<td></td>
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<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Right Leg</td>
<td>0.332 (0.132)</td>
<td>0.339 (0.153)</td>
<td>0.396 (0.119)</td>
<td>0.356 (0.147)</td>
<td>0.355 (0.127)</td>
<td>0.069</td>
<td>0.938</td>
<td>0.454</td>
</tr>
<tr>
<td>Left Leg</td>
<td>0.300 (0.117)</td>
<td>0.349 (0.134)</td>
<td>0.347 (0.124)</td>
<td>0.327 (0.133)</td>
<td>0.338 (0.120)</td>
<td>0.080</td>
<td>0.537</td>
<td>0.878</td>
</tr>
<tr>
<td><strong>CV ML COP velocity</strong></td>
<td></td>
<td></td>
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<td></td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Right Leg</td>
<td>0.283 (0.086)</td>
<td>0.311 (0.124)</td>
<td><strong>0.348</strong>&lt;sup&gt;a&lt;/sup&gt; (0.102)</td>
<td>0.311 (0.119)</td>
<td>0.317 (0.095)</td>
<td><strong>0.028</strong></td>
<td>0.734</td>
<td>0.549</td>
</tr>
<tr>
<td>Left Leg</td>
<td>0.272 (0.088)</td>
<td>0.310 (0.124)</td>
<td>0.296 (0.103)</td>
<td>0.287 (0.113)</td>
<td>0.298 (0.100)</td>
<td>0.123</td>
<td>0.491</td>
<td>0.659</td>
</tr>
</tbody>
</table>

<sup>a</sup>p < 0.05 vs. no load, <sup>b</sup>p < 0.05 vs. 20% BW bilateral load, <sup>*</sup>p < 0.05 even surface vs. uneven surface
Table 2. Mean and standard deviations for postural stability parameters for loaded limb stance vs. unloaded limb stance

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Loaded limb Mean (SD)</th>
<th>Unloaded limb Mean (SD)</th>
<th>p- value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Double stance ratio</td>
<td>0.489 (0.034)</td>
<td><strong>0.509</strong> &lt;sup&gt;*&lt;/sup&gt; (0.042)</td>
<td><strong>0.001</strong></td>
</tr>
<tr>
<td>ML COP excursion (mm)</td>
<td>10.09 (3.72)</td>
<td>11.01 (4.15)</td>
<td>0.149</td>
</tr>
<tr>
<td>ML COP velocity (mm/s)</td>
<td>25.4 (9.6)</td>
<td><strong>28.5</strong> &lt;sup&gt;*&lt;/sup&gt; (11.3)</td>
<td><strong>0.033</strong></td>
</tr>
<tr>
<td>ML TTC percentage</td>
<td>98.9 (1.1)</td>
<td>98.9 (0.9)</td>
<td>0.870</td>
</tr>
<tr>
<td>CV double stance ratio</td>
<td>0.053 (0.013)</td>
<td>0.055 (0.024)</td>
<td>0.676</td>
</tr>
<tr>
<td>CV ML COP excursion</td>
<td>0.396 (0.086)</td>
<td>0.347 (0.104)</td>
<td>0.082</td>
</tr>
<tr>
<td>CV ML COP velocity</td>
<td><strong>0.348</strong> &lt;sup&gt;*&lt;/sup&gt; (0.064)</td>
<td>0.296 (0.086)</td>
<td><strong>0.017</strong></td>
</tr>
</tbody>
</table>

* p < 0.05
Figure 1. Illustration of the three load conditions: no load (left), 20% BW bilateral load split between both sides of the body (center), and 20% body weight (BW) load on one side of the body (right).
Figure 2. Two walking surfaces: (a) uneven surface treadmill vs. (b) even surface treadmill.
Figure 3. Illustration of the rectangular boundary of the foot and the Time-to-Contact (TTC) calculation. \( v \) and \( a \) (COP velocity and acceleration) were calculated using the first central difference method.

\[
TTC_{(i)} = \frac{-v_{(i)} \pm \sqrt{v^2_{(i)} - 2a_{(i)}d_{(i)}}}{a_{(i)}}
\]

- \( d_{(i)} \): distance from COP to ML boundary
- \( v_{(i)} \): ML COP velocity
- \( a_{(i)} \): ML COP acceleration
- \( i \): each data point (100 Hz)
Figure 4. Illustration of TTC during walking: if TTC is less than remaining single stance time, then TTC saved. If TTC is greater than remaining single stance, then TTC is equal to remaining single stance time (no adjustment needed). TTC percentage was calculated by the ratio of TTC and remaining single stance (100% if no adjustment needed).
CHAPTER 5: SUMMARY

In my first study, we found that a 20% body weight (BW) unilateral load resulted in significantly higher peak L5/S1 lateral bending, hip abduction, and external moments that nearly all other loading conditions (no load, 10% BW bilateral load, 10% BW unilateral load, and 20% BW bilateral load) during stair ascent and descent. Therefore, we suggest potential benefits when carrying symmetric loads in order to decrease the frontal joint moments, particularly at the level of 20% BW loads. Furthermore, a 20% BW load is at or beyond a critical level where asymmetry results in a substantial increase in the frontal plane joint moments for stair negotiation.

In my second study, peak L5/S1 contralateral bending moments were significantly higher when carrying a 20% BW unilateral load as compared to a 20% BW bilateral load for both stair ascent and stair descent. In addition, peak L5/S1 contralateral bending moments were significantly higher during step one than for step two. Peak external knee varus and hip abduction moments were significantly higher in unloaded limb stance as compared to loaded limb stance, specifically when carrying a 20% BW unilateral load. Therefore, general load carriage recommendations include carrying less than 20% BW loads and splitting loads bilaterally when feasible.

In my third study, we found no significant differences in postural stability parameters when comparing a 20% BW unilateral load to a 20% BW bilateral load. Significant differences in coefficient of variation (CV) of double support ratio and medio-lateral (ML) center of pressure (COP) velocity were only found when comparing no load to the unilateral load, but not to the bilateral load. In addition, the uneven surface resulted in increased double stance ratio, ML COP excursion, ML COP velocity, and CV of double stance ratio as compared to the even surface. Therefore, walking on the uneven surface is more challenging in postural control as compared to walking on the even surface. During asymmetric load carriage, double stance ratio and ML COP
velocity were increased during the unloaded limb stance as compared to the loaded limb stance. Therefore, unilateral load carriage, walking on uneven surfaces, and balance on the unloaded leg may be of concern when considering postural stability and risk of falls or slipping.