ASYMMETRIC LOAD CARRYING WHILE WALKING ON A TREADMILL: GAIT KINEMATICS AND LOWER LIMB COORDINATION

By

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A THESIS PRESENTED TO THE GRADUATE SCHOOL
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To my family and friends, for all their support

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Bags (backpacks and single sling/messenger bags) have become a daily necessity for individuals of all ages. Carrying single strap bags induces asymmetrical loading and could be harmful to the human body due to altered postures while walking. The purpose of this study was to investigate the effect of different load carriage on gait kinematics and coordinative patterns in the lower extremities. Twenty-four university students participated in this study and walked on a treadmill at their preferred pace under three different load conditions: condition 1 (5% of body weight in messenger bags on each shoulder hanging vertically), condition 2 (10 % of body-weight in a messenger bag on one shoulder hanging vertically) and condition 3 (10 % of body-weight in a messenger bag on one shoulder with the bag draped across the trunk to opposite hip). We examined gait kinematics (stride length, cadence, step width and swing/stance ratio) and coordinative patterns in terms of intralimb and interlimb coupling utilizing Continuous Relative Phase analyses (CRP). CRP was evaluated over three interlimb (thigh-thigh, shank-shank, and foot –foot) and four intralimb couplings (thigh-shank and shank-foot in both legs). The results were analyzed by using repeated measures one-

way Analyses of Variance (ANOVA) and indicated improved gait kinematics (step with and swing/stance ratio) and coordinative patterns (thigh-shank, shank-foot and thigh-thigh couplings) during bilateral carriage compared to unilateral load carriage. Furthermore, asymmetric load carriage may have the potential to alter gait patterns and coordinative mechanisms in the lower extremities which may be associated with the risk of falling in taxing circumstances (e.g. slope, stairs, wet floors, etc.). Our results provide a picture of the adaptive mechanisms utilized by the locomotor system under external loading constraints.

CHAPTER 1
INTRODUCTION

People frequently carry loads using bags with shoulder straps. This allows individuals to transport a variety of items and still have their hands free. Bags (backpacks and single sling/messenger bags) have become a daily necessity for most people and are utilized by individuals of all ages. Backpacks are used by soldiers, hikers, and students. Students commonly use backpacks to carry their books, computers, sports equipment, and other items on a daily basis (Negrini et al., 1999; Grimmer et al., 1999; Grimmer & Williams, 2000; Whittfield et al., 2001). In recent years, single strap bags have become more popular with students although they have been used extensively by bicycle couriers for years and even by riders of the Pony Express as early as 1860.

The majority of previous research related to carrying loads has been related to conventionally worn backpacks. In fact, carrying symmetrically loaded backpacks using both straps influences gait kinematics (Birrell & Haslam, 2009), posture and heartrate (Simpson et al., 2011a), muscle activity patterns (Simpson et al., 2011b), balance (Pau et al., 2011; May et al., 2009), and even decision making (May et al., 2009). Wearing backpacks has also been suggested as a possible treatment for osteoporosis (Wendlova, 2011) and camptocormia (Sakas et al., 2010). However, there is a relative dearth of research concerning single strap bags that are associated with requisite asymmetric loading.

Asymmetric loading occurs when load center of mass is displaced laterally in the frontal plane (DeVita et al., 1991) which is observed when carrying a single sling or messenger bag. This displacement of load center from body mid-line could affect the

gait pattern and cause back pain, as well as, injury of lower limbs and joints. Although carrying backpacks over both shoulders could be the best way to prevent back pain (Macias et al., 2008; Korovessis et al., 2005), students also choose asymmetrical load carriage instead of symmetrical for a variety of reasons ranging from convenience to fashion. In other words, carrying single strap bags are not conductive to the health of individuals' back, shoulder, or lower extremity and joints (Pascoe et al., 1997; Negrini et al., 2007; Macias et al., 2008; Matsuo et al., 2008).

Students induce an asymmetric load by carrying their backpack with one strap on one shoulder instead of both shoulders, concentrating the load on one side of the body. This loading can place considerable stress and strain on the shoulder muscles (Marras et al., 1997). It also seems that by carrying a single strap on one shoulder, individuals must bend their trunk contralaterally to address the demand for dynamic balance. Accordingly, increased trunk flexion toward the contralateral side (DeVita et al., 1991; Pascoe et al., 1997) due to asymmetric load-carrying causes an increase in hip abduction torque (DeVita et al., 1991; Matsuo et al., 2008). This indicates that asymmetric loading alters dynamic balance when wearing a backpack with one strap compared to both straps (Pascoe et al., 1997). In addition, the increased lateral bending caused by asymmetric loading increases spinal stress (DeVita et al., 1991; Pascoe et al., 1997; Legg & Cruz, 2004). The alterations of Contralateral trunk flexion during walking may cause scoliosis or other permanent physical problems (Roaf, 1960; Hawes & O'brien, 2006). As a result, increased bending stress on the trunk and spine may exacerbate any abnormalities in the biomechanics of locomotion. These alterations suggest that single strap bags may be a contributor to abnormal posture and motion of

the lower extremities in gait. Interestingly, the loads and mechanical adaptations experienced when carrying loads tend to take place in the frontal plane. However, the goal of load carriage during gait is to transport various items in a forward (sagittal plane) direction. Therefore, most of the parameters examined in the current study were sagittal plane variables. In addition, sagittal plane movements of the lower extremities are typically the most meaningful measures compared to the other two planes in terms of moving the body forward.

Physiological responses of single strap bags on gait measures can be seen carrying golf bags. The muscle activation to maintain upright position when carrying the single-strap of a golf bag on one shoulder, compared to double strap bags, may cause higher demand on metabolic cost that induces fatigue (Ikeda et al., 2008). Asymmetric load carriage is associated with higher oxygen uptake (Ikeda et al., 2008; Legg et al., 1992) and heart rate (Ikeda et al., 2008) than backpack load carriage and could increase muscle activation (Neumann et al., 1996) and fatigue. Fatigue caused by single strap athletic bags could be one factor that may change posture and gait pattern.

In attempts to study posture and gait, previous researchers have modeled lower extremity movement as a pendulum, with the segments of the leg oscillating during the gait cycle. Coordinative structures have been useful tools to resolve and map each segment pattern during gait. In movement science, examining how two oscillators are coupled together during a limit cycle has been an important concern since human movement involves coupling of multiple degrees of freedom in the human body. Moreover, previous researchers have attempted to quantify coordinative patterns of two oscillating segments during gait cycles by utilizing several methods in terms of

dynamical system theory (e.g. continuous relative phase, vector coding, and cross-correlations). In this study, we used continuous relative phase to quantify and analyze limb coordination of the lower extremities during the gait cycle. More specifically, we adopted CRP to investigate effect of carrying messenger bag that affect inter limb and intra limb coordination during the gait.

In sum, there has been a noticeable increase in single strap bags sales which would indicate that more people are carrying them (Vicky, 2001). With the research on numerous type of bags (schoolbags, military bags, athletic bags, and messenger bags) showing that indeed, bags have a direct effect on the movement of the human body, there seems to be a lack of research on other types of bags such as single strap bags. In everyday life, bags can be considered necessities and therefore must be considered if the movement is to be fully understood. A few researchers have investigated how asymmetric load carrying affects the biomechanics of human movement. However, few data exist regarding asymmetric load carriage and gait kinematics in terms of coordination patterns of the lower extremities. Therefore, we attempted to show evidence indicating that carrying single strap bags induces abnormal coordination patterns in the lower body while walking.

Specific Aims

The aims of the research were 1) to investigate gait kinematics in response to three different loading conditions and 2) to evaluate lower extremity limb coordination in response to different loading conditions in university students during treadmill walking at a self-selected pace. Baseline conditions were not included for gait kinematics because the authors' focus was to measure gait alterations in response to bilateral and unilateral load carriage.

Hypotheses

The following general hypotheses and specific sub-hypotheses were tested.

1. Gait pattern would be altered when carrying two messenger bags (one on each shoulder) compared to a messenger bag on one shoulder.

    A. Stride length would decrease and cadence would increase during symmetric loading compared to asymmetrical loading.

    B. Step width would decrease during symmetric loading compared to asymmetric loading.

    C. Swing/stance ratio would decrease on the loaded side and increase during symmetric loading compared to asymmetric loading.

2. Lower limb coordination (inter and intra limb coordination) would be altered when carrying two messenger bags (one on each shoulder) compared to a messenger bag on one shoulder.

    A. RMS differences for intralimb (thigh-shank and shank-foot couplings) and interlimb (thigh-thigh, shank-shank and foot –foot couplings) would decrease during symmetric loading compared to asymmetric loading.

    B. Cross-correlation coefficient values for intralimb and interlimb couplings would increase during symmetric loading compared to asymmetric loading.

CHAPTER 2
LITERATURE REVIEW

Single strap bags have become more popular with students in recent years, but have been linked to altered gait kinematics (Pascoe et al., 1997). The proposed study will be undertaken to examine the kinematic changes that may lead to variations in the lower extremity movements and coordination. This chapter will be divided into four sections. Section one will describe human locomotion and the phases of gait. Section two will describe limb coordination during human gait. Section three will describe a dynamical systems' approach for analyzing the coordination of human movement. Finally, section four will describe the influence that asymmetrical load-carrying has on human locomotion as well as coordination patterns of lower extremity.

Human Locomotion and the Phases of Gait

Human gait is very complicated, involving continuous coordinated bipedal locomotion. As a person moves forward, one of the lower limbs serves as a mobile source of support while the other limb moves forward to become the new support site in front of the current support site. To transfer body weight from one limb to the other, both are alternately placed on the ground. This series of leg motions is called a "gait cycle"; each gait cycle begins with ground contact of the heel and ends with the following heel contact of the same limb (Figure 2-1). Several approaches have been proposed to fully investigate the human gait cycle and numerous terminologies to define specific human walking events have been introduced. In this chapter, we will look over the descriptive phrasal terminology yielded from previous observational and kinematic analyses of normal gait (Perry 1992; Sutherland et al., 1994).

Since this cycle of events is repeated by each limb with reciprocal motion timing to the following part, there is no specific ending or starting point. However, for gait analysis, an initial point of gait will be used to define the beginning of each gait cycle. Each gait cycle is divided into two periods (stance and swing) based on heel contact points of reciprocal foot motion (Figure 2-1). The swing phase is the time period in which the foot is not in contact with the ground, starting with the point of toe off and ending with the point of heel contact. The stance phase of the gait cycle is the time period in which the foot is in contact with the ground. The stance phase begins with heel contact, and ends with the point of toe-off.

The stance and swing phases are subdivided into three different intervals based on the sequence of heel contact. The stance phase is subdivided into three smaller phases: initial double support, single support, and terminal double support. The gait cycle begins with initial double support phase when both feet are on the ground after initial contact. When the opposite foot is lifted and starts to swing, single support begins. In this phase, one leg is in contact with the ground to support the entire body weight, while the other leg swings forward. When the swing leg begins a new heel contact, the single support phase ends and terminal double support phase begins. Terminal double support continues until the original stance limb is lifted to make a new swing phase. The swing phase is also subdivided into three phases: initial swing, mid swing, and terminal swing. The initial swing phase is the time from when the swing foot leaves the ground to when it crosses the stance foot. In this phase, the foot is off the ground and accelerating forward. Next, the mid-swing phase begins when the swing limb is opposite the stance foot and ends when the tibia is vertical. In the mid-swing

phase, the swing foot moves forward and overtakes the limb in stance. Terminal swing is from the time when the tibia is vertical until the foot makes contact with the ground. Finally, the decelerating swing limb completes swing phase in preparation for the stance phase.

In normal gait, the stance phase accounts for approximately 60% and the swing phase occupies around 40% of the total gait cycle although the precise duration of the intervals is altered according to the individual's walking velocity (Mann, 1982). Also, the time of the stance and swing phases of the gait cycle are inversely related to the walking velocity in that both the stance and swing phases decrease in duration as velocity increases (Mann, 1982). Likewise, when walking velocity slows, the times of the stance and swing phases become longer.

Over the past several decades, numerous *temporal* technical terms describing gait have been produced to help us scrutinize human gait. Moreover, a variety of gait parameters has been codified to analyze and quantify gait. The terminology related to human walking began with descriptive phrases obtained as a result of observational and kinematic analysis of normal participants (Perry, 1992). For example, stride is the time period from the heel strike of one foot to the following heel strike of the same foot. Steps, regarded as the timing between two limbs, are indicated by the time period of initial contacts of both feet. That is, a step is the time period from the heel strike of one foot to the heel strike of the other foot. Two steps have been identified in one gait stride, since the other initial contact begins at the midpoint of one stride (Perry, 1992, Figure 2-2).

The evaluation of walking is the primary tool for describing human gait patterns. In the field of biomechanics, gait analysis is a method by which modern technologies are

used to incorporate information from a number of inputs to illustrate and analyze the dynamics of gait (Perry, 1992). This basic approach to evaluating gait will be the focus of the current research. It will provide evidence which will be used to address the specific aim.

Limb Coordination

As previously stated, human gait is a complicated movement that includes cycles and coordination of each segment. In bipedal locomotion, limb coordination is crucially important and must be altered according to demands of varying external circumstances (Reisman et al., 2005). To adapt to the demands of various environments, specific coordinative patterns occur both within limb segments and between limbs. For example, coordination of oscillating segments may change based on external restraints, such as surface properties, or internal restraints, such as a neurological disease (Haddad et al., 2005). Functional gait demands flexible limb movements to accommodate for different external constraints (Reisman et al., 2005). Inter-segmental coordination demands complicated interaction between the motor output of the neuromuscular system, biomechanical factors, environmental factors, and task constraints (Higgins, 1985; Kugler & Turvey, 1987). In response to the demands of a specific task, muscles must be used in combination to bring about particular movements in several joints (Turvey, 1990). Therefore, movement coordination may involve both muscular action and interactions at each joint. This approach to human movement may allow scientists to implement important insight in their investigation of the neurological system. Thus, human movement coordination has been defined as the neurobiological system's ability to generate complex movement incorporating several segments or joints (Forner-Cordero et al., 2005). Coordinative inter-joint and inter-segmental movements are a

strategy of the central nervous system to control movement and maintain stability during human locomotion (Kurz et al., 2004). Different coordination patterns have been observed for various neurological disorders (i.e. hemiparesis, parkinson's disease, stroke). In hemiparetic gait, an overactivated rectus femoris during swing phase was associated with changing thigh-shank coordination (Forner-Cordero et al., 2005). Other research suggests that asymmetric movement coordination in parkinson’s disease may be involved in freezing of gait (Plotnik et al., 2005). Therefore, coordinative inter-segmental movement results from individual muscles and neuropathways (Stergiou, 2004). Effective organization and modeling of gait coordination has been an important focus for many researchers. Traditionally, biomechanical tools have been utilized to analyze human movement patterns. However, scientists need novel approaches to gain a more advanced understanding of the organization of human movement. The two components of this approach are the multifactorial and nonlinear relationships of the gait pattern (Stergiou, 2004). Moreover, mapping coordination patterns in limbs during gait provides a clear mechanism of the neuromuscular system in low-dimensional terms (Kurz et al., 2004). In a biomechanical system, to organize a coordinative model, researchers have reduced the high available degrees of freedom in segmental movement (Hamill et al., 1999). The concept of coordinative structure comes from Bernstein’s work that indentified problems regarding the numerous degrees of freedom and made efforts to simplify the many degrees of freedom in segmental movements and reduce independent variables requiring control (Bernstein 1967; Turvey, 1990). In conclusion, previous researchers studying gait coordination have been focused on organizing the system for functional movement pattern, reducing numerous independent

variables and, summarizing the relationship between various components during gait. Examining limb coordination patterns under external constraints (i.e. asymmetric load) during gait may provide insight into specific control mechanisms and human movement adaptations under various environmental conditions.

Continuous Relative Phase

Researchers have used different nonlinear dynamic techniques to investigate variability in human movement based on the dynamical systems theory (Miller et al., 2010). One prevalent dynamic systems analysis for studying coordination of segmental movement can be evaluated through continuous relative phase (CRP). This measure has been used to quantify the coordination between different body segments in several activities (Varlet & Richardson, 2011). This approach to investigating the coordinative pattern of gait is to recreate it as a dynamic system and study its stabilizing features. Moreover, CRP indicates the aspects of interacting segments during the entire movement cycle. Therefore, the relative phases of several oscillating segments can be measured to quantify segmental coordination and evaluate movement patterns.

Prior to calculation for CRP, each segment angle (Figure 2-3) is normalized for each trial using the following equation:

$$\text{Angle}: \theta_{N_i} = \frac{2 * [\theta_i - (\theta_{max} + \theta_{min})]}{\theta_{max} - \theta_{min}}$$

$$\theta : \text{angle}$$

$$\theta_N : \text{normalized angle}$$

$$i : \text{data point}$$

Also, angular velocity is normalized the following equation:

$$\text{angle velocity} : \omega_{N_i} = \frac{\omega_i}{\max\{\max(\omega_i), \min(-\omega_i)}$$

$$\omega : \text{angle velocity}$$

$$\omega_N : \text{normalized angle velocity}$$

$$i : \text{data point}$$

This is done by plotting the posing of a segment angle versus the angular velocity of that segment in the “phase plot” (Hamill et al., 1999). The phase angles (Φ) are obtained by calculating four-quadrant arctangent of the ratio of angular velocity by angular position:

$$\varphi_i = \tan^{-1}\left[\frac{\theta_{N_i}}{\omega_{N_i}}\right] \varphi_i = \tan^{-1}\left[\frac{\theta_{N_i}}{\omega_{N_i}}\right]$$

$$\varphi : \text{phase angle}$$

$$i : \text{data point}$$

Then the continuous relative phase is calculated by subtracting the phase angle of the proximal segment from that of the distal segment for a specific point during the gait cycle.

$$\text{CRP} = \varphi_{\text{proximal}} - \varphi_{\text{distal}}$$

Where continuous relative is the relative phase angle between the distal and proximal segment, Φdistal segment (i.e., shank) is the phase angle of the distal segment, and Φproximal segment (i.e., thigh) is the proximal segment. *The main strength of the CRP measure is that it compresses four variables* (i.e., proximal and distal segments’ angles and angular velocities). When the CRP is near 0°, the respective segments are in-phase, while 180°of the CRP indicates that both segments are anti-phase (Haken et al.,1985; Kelso et al., 1986; Scholz & Kelso, 1989; Diedrich & Warren, 1995). Positive relative values indicate that the distal segment is ahead of the proximal segment, and negative

values indicate that the proximal segment is ahead in phase space (Clark & Phillips, 1993; Barela et al., 2000). The slope of the relative phase curve configuration indicates which segment is moving faster during the period of the gait cycle (Barela et al., 2000). A positive slope indicates that distal segment is moving faster in phase space, while a negative slope indicates that the proximal segment is move faster in phase space. Comparing the phase angles of two segments using relative phase provides insight of normal and pathological gait patterns (Kurz et al., 2004).

Influence of Asymmetrical Load-Carrying on Gait

Because bag conditions are the main external constraints for students during gait, the choice of bags can be a crucial factor in alterations of gait patterns. It seems that by carrying single strap bags, abnormal gait patterns may be produced that have the potential for inducing injuries in the lower extremity. Much of the research available in the biomechanics field on asymmetrical gait has shown a detrimental influence on the human body. Specifically, physical stress associated with carrying bags on one shoulder altered the posture and gait pattern of children (Pascoe et al., 1997). Pascoe and colleagues studied sixty-one youths, without bag, carrying a one-strap book bag, a two-strap bag, and a one-strap athletic bag. In this study, the impact of carrying book bags on static posture and gait kinematics was examined. This study utilized a two-phase approach. In phase one, sixty-one students who transport their school materials had a survey that provided descriptive and anthropometric characteristics, book bag weight, typical bag carriage, and associated physical symptoms. In phase two, ten participants were chosen for kinematic analysis based on the survey and underwent dynamic walking at a self-determined pace. A 7.7kg (17% of body weight) book bag weight was used in this study. A video analysis system was employed for digitizing the movements

for kinematic analysis. The results of the study indicated that the load of the book bags resulted in a decrease in stride length and an increase in stride frequency and greater angular motion of the head and the trunk was observed with the athletic bag trial. Moreover, they showed carrying bags on one shoulder caused different movement strategies such as lateral spinal bending and shoulder elevation while walking. They also suggested that carrying backpacks promoted a significant forward lean of the head and trunk compared to normal walking or one-strap athletic bags. In sum, they showed clear evidence that carrying backpacks or messenger bags altered the posture and gait of youths. However they did not evaluate the gait patterns of lower extremities related to these abnormal postures.

Another study focusing on the effects of carrying an asymmetric sidepack on the frontal plane joint moments in both lower extremities and in the L5/S1 joint during walking was performed by Devita and Colleagues (1991). Five healthy males were recruited for this study and required to walk under three load conditions: no load, a load equivalent to 10% of the subject's body weight, and a load equivalent to 20% of the subject's body weight. The participants were instructed to carry the pack on the left shoulder while the subject’s left hand held the front of the strap. In this study, the experiment was designed to obtain three-dimensional ground reaction force data and frontal and sagittal plane film records for deriving lower extremity joint moments in both plane and frontal plane L5/S1 moments. The results of this study showed that asymmetric loads produced unbalanced frontal plane moments, large changes in right limb moments and smaller changes in left limb moments and caused L5/S1 moment dominance to shift to the right side during left and right single support phase. The

authors conclude that since the asymmetric load caused unilateral and unbalanced use of trunk muscles on the non-loaded body side for both stance phases, asymmetric load carrying can be a greater risk factor for injury than symmetric carrying (Devita et al., 1991). Although they also showed different movement patterns such as unbalanced hip and knee joint moments that may alter limb coordination, they did not report coordination patterns in the lower body during asymmetric load carrying.

A separate study provided clear evidence that carrying single strap mailbags alter the kinematics of the spine and induce stature loss (Fowler et al., 2006). They used a specific mail bag (17.5% body weight load) to simulate the task of the postal worker and attempt to quantify the effect of asymmetric load carriage on spinal kinematics and stature loss. Thirteen markers were attached on the participant's back and one marker over each shoe. The changes in stature measure were assessed with a stadiometer. The results of this study indicated increased forward leaning and bending of the spine while carrying the messenger bag during walking. More specifically, they showed up to five degrees of forward flexion in the thoracic spine in the sagittal plane and up to a 12° increase in the lumbar region in the frontal plane. Stature loss was evident in the loaded condition, producing a spinal shrinkage greater than the unloaded condition (12.1±1.2mm loaded, 5.75 ±1.1mm unloaded). These findings provided strong evidence that carrying asymmetric loads induces abnormal postures related to spinal configuration during gait.

In addition, Hadded and colleagues (2005) performed a study on the coordinative pattern in lower extremity to asymmetrical loading during walking. In this study twelve healthy individuals were recruited to walk on a treadmill at their preferred walking speed.

Four different unilateral loads (0.9, 1.8, 2.7, 3.6, 4.5kg) for each lower leg were worn 2.5cm above the lateral malleolus utilizing a custom-made leg loading device. The researchers hypothesized that this adaptation would alter interlimb coordination while intralimb coordination would remain invariant. Continuous relative phase (CRP) was employed to evaluate changes in limb coordination patterns. Changes in coordinative patterns were quantified using both cross-correlation and root-mean-square (RMS) difference measures. The result of this study indicated that increases in leg loads resulted in RMS changes in the interlimb level of the loaded leg. At the interlimb level, significant differences were found in both cross correlation and RMS measures (Haddad et al., 2004). The authors of this study indicated that various adaptations in lower limb coordination appear at both the intralimb and interlimb levels in response to unilateral load changes. They suggested changes at the intralimb coordination level were greater than the changes in interlimb coordination, in both RMS and cross-correlation measures. More specifically, at the intralimb level (the thigh-shank and shank-foot couplings), the RMS difference in the loaded leg increased significantly (indicating out of phase movement) during both stance and swing phases. At the interlimb level, the RMS differences significantly increased in all three interlimb couplings (thigh-thigh, shank-shank, and foot-foot). Also, the cross-correlation coefficient systematically decreased with greater loads. However they used a custom-made leg loading device that was placed 2.5cm above the lateral malleolus. This load condition at the lower leg may cause limited leg movement due to load itself and is difficult to apply to real life. In the current study, single strap bags carried on one shoulder were used for enhancing potential asymmetries instead of this leg load.

Yet another study investigated balance during asymmetric load-carrying and how asymmetric loading affects lower limb coordination while walking. Five young women and six elderly women walked under three different conditions: without a bag, with a 3kg hand-held bag, and with an 8kg hand-held bag. The researchers also used CRP measures to analyze lower limb coordination and CRP means for each load condition were used to calculate the cross-correlation coefficients. The results of this study showed that the mean of cross correlation coefficients for both interlimb and intralimb level were close to 1.0 and each participant showed the same coordination pattern. The author concluded that limb coordination is maintained under external constraints such as asymmetric loads, thus hand-held load carrying minimally affected the coordinative patterns in lower limbs (Matsuo et al., 2008). This study used 3kg and 8kg women's handbags as its load conditions but we used messenger bags to evaluate lower leg movement in response to the diverse load conditions that are usually encountered by university students.

Several studies have shown that asymmetrical load-carrying creates issues with kinematics and coordinative patterns in lower extremity. Even though previous research examined gait kinematic changes and coordinative patterns in lower extremity under various loads conditions, little research has been done to show exactly how carrying a single strap bag, compared to carrying a backpack, affects gait kinematics and lower limb coordination for university students. Therefore it is important that more research be conducted to investigate the effect of single strap bag carrying on coordination patterns in lower extremity, providing insight into circumstances with a potential for injury. The present study will be an attempt to simulate real life loading conditions utilizing different

types of bags. Furthermore, the evaluation of coordination patterns may reveal insight regarding neuromuscular mechanisms related to gait.

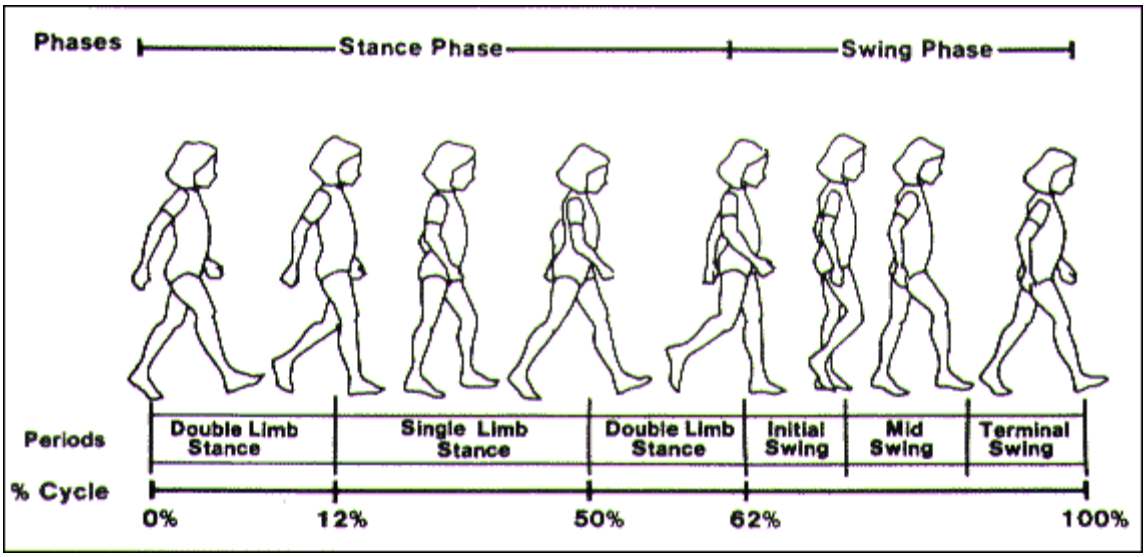


Figure 2-1. Gait phases and subdivisions of stance and swing phase (Kirtley, 2008).

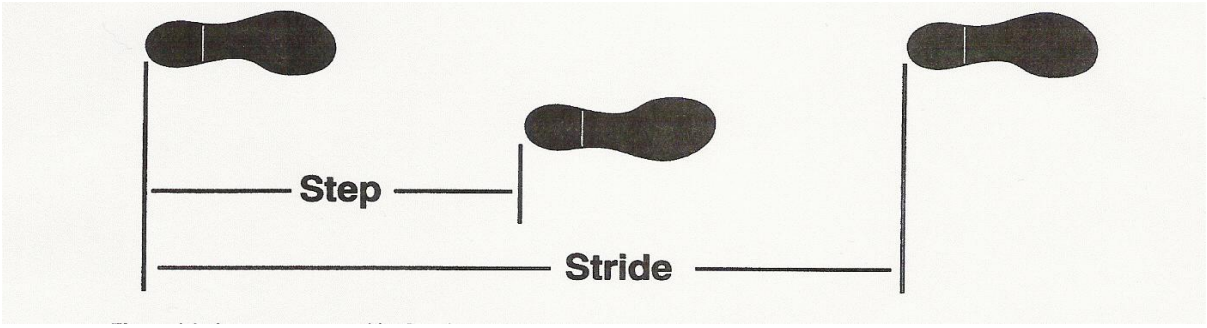
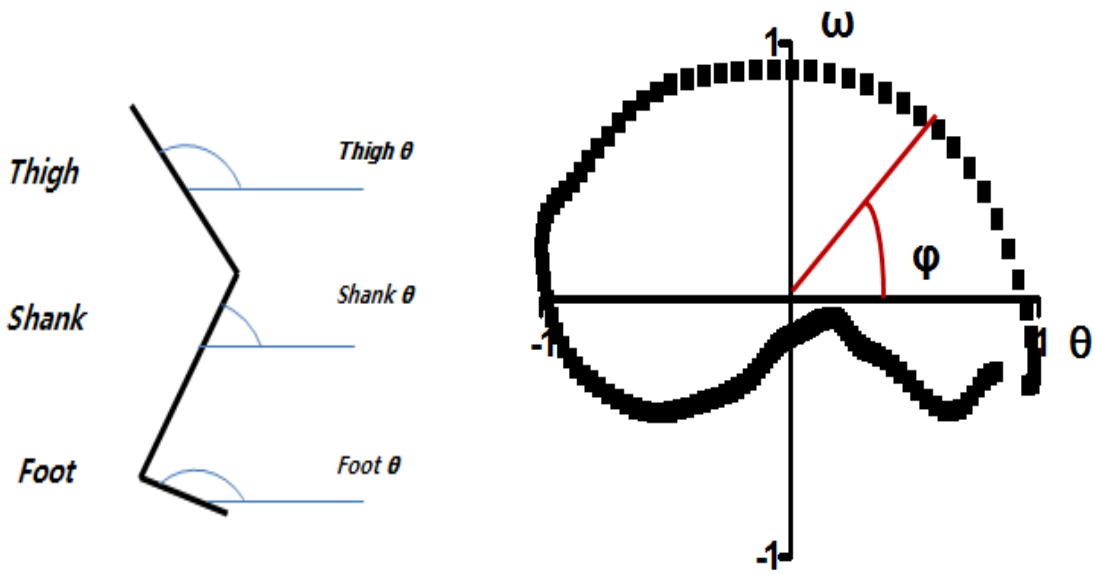


Figure 2-2. Stride and step: A step is defined as the interval between heel contact of opposite feet. Two steps compose one gait stride.



(A) (B)

Figure 2-3. Illustration of each segment angle and phase plot (A) Illustration of each segmental angle: thigh, shank, and foot angles in sagittal plane. (B) Phase plot illustrating phase angle based on angular displacement versus angular velocity over one gait cycle. Calculation of Phase angle(φ) of thigh and shank was obtained from arctangent function of angular velocity(ω)/ angular displacement (θ).

CHAPTER 3
METHODS

Participants

Twenty-four healthy students with an age range of 18 to 30 (12 males and 12 female; age 21.6±3.6 years; height 170.9±8.5; weight 67.2±12.5kg) participated in this research. All participants were recruited from the University of Florida and surrounding community by word of mouth. All were free of any pathology that would prevent them from walking on a treadmill. Moreover, individuals were excluded if they had current back or neck pain, any joint pain, history of arm pain, or spine pain in the previous three months or if they were not healthy enough to handle the demands of the tests in this study. Prior to participating in the study, each subject read and signed an informed consent form approved by the university's institutional review board. Twenty-two participants were right-handed and during conditions 2 and 3 preferred to carry the messenger bag on the right shoulder, while the other two left-handed individuals preferred to carry the bag on left shoulder. Though both sides were loaded during condition 1, the "loaded" and "unloaded" labels in the remaining portions of this document refer to the sides which are loaded or unloaded during conditions 2 and 3, respectively.

Instrumentation and Task

Two single strap bags (Figure 3-1) were utilized in this research to create three different load conditions: Two single strap bags, one on each shoulder and hanging down vertically (0% of body weight), two 5% body-weight single strap bags with one on the right shoulder and one on the left (hanging vertically), and 10% body-weight single strap bags in different positions while walking on a treadmill. The total weight carried in

the bags was normalized according to each subject's body weight (10%BW) based on existing literature as indicated in the following sentences. Previous studies using various loads (10-20%) have shown the diverse effect on the human body during gait (Devita et al., 1991; Fowler et al., 2005; Korovessis et al, 2005; Zultowski et al, 2008). In fact, the bag load for college-age students has been recommended to not exceed 10-15% of the individual's body weight to prevent injuries and pains in the body (AN et al., 2010). Thus, the load for the current study was determined based on these data and guidelines. Furthermore, the messenger bags were positioned approximately 10 cm below the ASIS makers across the conditions.

A Vicon motion analysis system with 7 high-resolution cameras (Vicon Nexus, Oxford, UK) was used to collect three-dimensional kinematic data during each testing condition. The cameras were positioned to record movement in the three cardinal planes (frontal, sagittal, and transverse) during the gait cycle. Kinematic data were captured at 120Hz. Calipers were used to record participants' lower extremity anthropometric measures and joint center locations were calculated from the static trial. All participants walked on an instrumented treadmill (Figure 3-2) for each load condition. The treadmill was built on two forceplates that allowed for measuring continuous ground force data during gait. These data were used for identifying gait events (toe-off and heel strike).

Procedures

After meeting the qualifications for participation, the body weights of participants were measured to determine appropriate 10% body weight loads to be carried in the different bag conditions. Participants were then prepared for retro-reflective marker placement by changing into tight fitting shorts and shirts. Sixteen retro-reflective

markers were placed on the lower extremity over bony landmarks according to the Vicon Plug-in-Gait marker system (Figure 3-3). All participants were instructed to walk at a self-selected speed on the instrumented treadmill. To determine the "self-selected speed" of each subject, the belt speed was initiated at 0.5 m/s. The speed was gradually increased in increments of 0.1m/s until the participant signaled that his or her preferred speed has been reached. Participants' lower extremity movements were captured digitally as they walked under different loading conditions. More specifically, participants underwent testing in four different load conditions: no bag, carrying two messenger bags on both shoulders with a 10%BW load, carrying the messenger bag on one shoulder hanging vertically down to the hip with a 10%BW load, and carrying the messenger bag on one shoulder draped across the trunk to opposite hip with a 10%BW load (Figure 3-1). In condition 2 and 3, the messenger bags were worn on the side of the preferred shoulder. They initially were asked to walk on a treadmill at their preferred pace for five minutes in each condition, with the bag order randomly assigned (by using a randomizing table). Thus, a total of four trials were completed (20 minutes of treadmill walking). Each participant was allowed to rest as much as necessary between trials. We collected data for the last one minute of each condition because of the large magnitude of data necessary for processing. Thus, around 50 strides in each condition were recorded.

Data Processing

As alluded to previously, three-dimensional lower extremity gait analysis was performed on all participants. Kinematic data were processed and exported using the Vicon nexus 1.7.1 software program (Vicon Nexus, Oxford, UK). Marker trajectories were filtered using a fourth-order Butterworth filter, with a 10Hz cutoff frequency. Then,

kinetic data were used to identify the stance and swing phases of the gait cycle based on each gait event (foot strike and toe off). Accordingly, the gait cycle was divided into two phases: stance and swing.

Gait Parameters: Hypothesis 1

We examined gait kinematics (stride length, cadence, step width and swing/stance ratio) while walking on a treadmill. Stride length was defined as the displacement of the ankle marker along the walking axis during the time period from the heel strike to toe off in the same foot (Reisman et al., 2005). Cadence was defined as the number of steps during one minute. Step width was defined as the mediolateral distance from the swing limb heel marker to the stance limb heel marker at each heel strike. Swing/stance ratio was defined as the ratio between swing and stance time over a gait cycle. Around 50 gait cycles were evaluated for the gait kinematics using a custom Matlab code.

Limb Coordination: Hypothesis 2

Three segmental angles (thigh, shank, and foot), as well as, angular velocities were assessed from raw kinematic data exported from the Vicon system (Figure 3-3). Continuous relative phase (CRP) analyses were performed using Matlab. We focused the kinematic analysis on CRP measures during the stance and swing phases of the gait cycle, as well as, the gait cycle as a whole. Segmental angular velocities were calculated from segmental angles in the sagittal plane utilizing the central difference method. These data were then used to calculate phase angles from a phase plot, using the arctangent of angular velocity / angular displacement at each data point.

For the intralimb couplings, CRP was calculated by subtracting the phase angle of the proximal segment from that of the distal segment for each data point. More specifically, the intralimb couplings were the thigh-shank and shank-foot in both limbs. For the

interlimb coupling, CRP was calculated by taking the difference between the phase angles of both segments for each data point. The interlimb couplings were the thigh-thigh, shank-shank and foot-foot. These procedures were described in more detail in Chapter 2.

Thus, CRP was evaluated over three interlimb (thigh-thigh, shank-shank, and foot –foot) and four intralimb couplings (thigh-shank and shank-foot in both legs) during the gait cycle, as well as the swing and stance phases of the gait cycle. Since each trial had a different length cycle, the data were processed using a linear length normalization that allows the different trials to have equal data lengths for each gait cycle. In interlimb coupling, the time series of one limb was time shifted to match left and right foot strikes. Coordination patterns were quantified utilizing cross-correlation coefficient (CCC) and root-mean-square (RMS) techniques. CCC was assessed by comparing the average CRP in each load condition to the average CRP in the no-bag baseline condition during gait cycle for interlimb and intralimb couplings. RMS difference was also evaluated by comparing the average CRP in each load condition to the average CRP in the no bag baseline condition. While the CCC measure indicates changes in the spatio-temporal evolution of CRP patterns, RMS measures show information about the magnitude differences in relative phase between the patterns (Haddad et al., 2004).

Statistical Analyses

Statistical analyses were performed using the SPSS® statistics (version 20; SPSS Inc., Chicago, IL, USA). The effect of the different loading conditions on gait kinematics (stride length, cadence, step width, and swing/stance ratio) and limb coordination parameters were analyzed by using repeated measures one-way Analyses of Variance (ANOVA). In addition, the influence of loading was further investigated for swing/stance

ratio and intralimb coupling by performing the statistical tests on the loaded and unloaded limbs separately. Bonferroni's posthoc procedure was used when appropriate. The level of statistical significance for all tests was set at $P<0.05$.

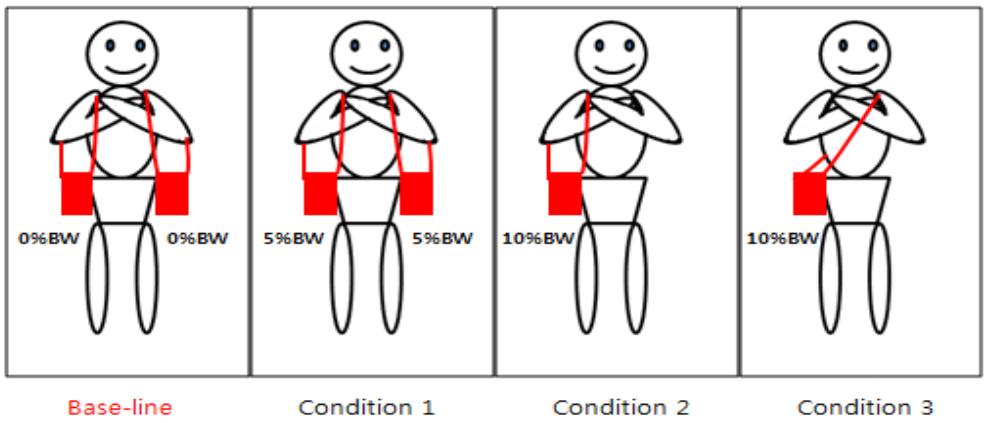


Figure 3-1. Illustration of four different load conditions. Baseline (no load with one messenger bag on each shoulder hanging vertically down to the hip), condition 1 (5% of body weight in messenger bags on each shoulder hanging vertically), condition 2 (10 % of body-weight messenger bag on one shoulder hanging vertically) and condition 3 (10 % of body-weight messenger bag on one shoulder with the bag draped across the trunk to opposite hip).

Figure 3-2. Bertec Instrumented Treadmill (Dual belt treadmill, Bertec, Columbs, Ohio, USA).

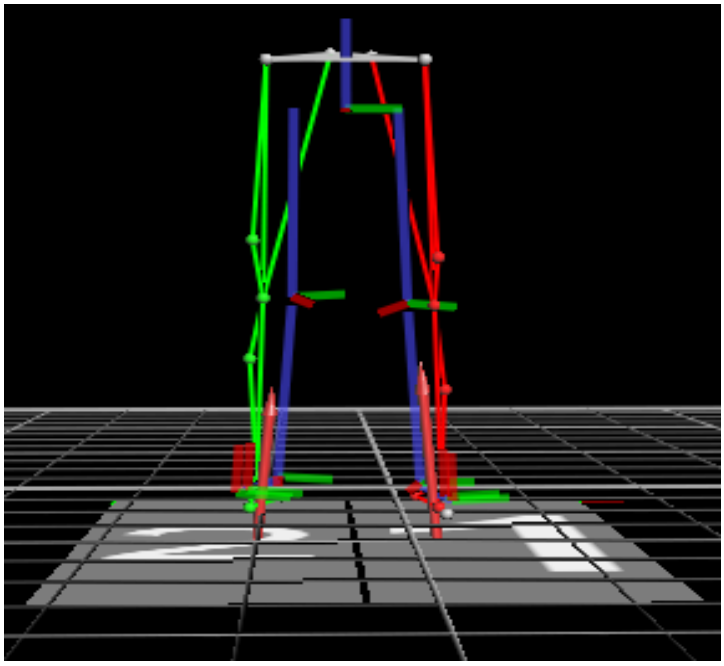


Figure 3-3. Illustration of plug-in gait model in lower extremity with 16 lower extremity markers.

CHAPTER 4
RESULTS

Gait Parameters: Hypothesis 1

Initially, we evaluated the effect of different loading conditions on gait kinematics via stride length, cadence, step width and stance/swing ratio. A main effect for stride length was detected ($F(2, 46) = 5.418$, $P = .008$). Follow-up testing revealed that stride length during condition 2 was shorter than condition 3 ($p = .006$, Figure 4-1). No other significant differences between conditions were observed ($P > .05$). Also, cadence varied across conditions ($F(2, 46) = 5.902$, $P = .005$). More specifically, cadence during condition 2 was significantly increased compared to condition 3 ($P = .005$; Figure 4-1). No difference in cadence was observed in condition 1 versus 2 or condition 1 compared to 3 ($P > .05$). A step width main effect was observed ($F(2, 46) = 14.481$, $P < .001$). Post-hoc analysis revealed that the step widths for condition 1 compared to conditions 2 and 3 were significantly decreased ($P = .002$, $P < .001$; Figure 4-1). There was no other significant difference between conditions for step width ($P > .05$). Finally, the swing/stance ratio was assessed in both the loaded and unloaded sides over a gait cycle. As mentioned previously, though both sides were loaded during condition 1, the “loaded” and “unloaded” labels refer to the sides which were loaded or unloaded during conditions 2 and 3, respectively. A significant main effect was observed in the loaded limb ($F(2, 46) = 14.274$, $P < .001$). In particular, the swing/ stance ratio on the loaded side during condition 1 was increased compared to conditions 2 and 3 ($P < .001$ and $P = .001$, respectively; Figure 4-2). No significant difference was observed between condition 2 and 3 ($P > .05$). Additionally, a main effect of swing/stance ratio on the unloaded side was noted ($F(2, 46) = 5.738$, $P = .006$). Subsequent testing indicated

that the swing/stance ratios during condition 1 were significantly decreased compared to conditions 2 and 3 (P = .014, P = .033; Figure 4-2). No difference was observed between conditions 2 and 3 (P >.05). A summary of these comparisons appears in Table 4-1.

Limb Coordination: Hypothesis 2

Intralimb Coordination

Intralimb couplings (thigh-shank and shank-foot) were examined for the loaded and unloaded side during the stance and swing phase to investigate the mechanism of coordinative patterns in the lower extremities over a given period. Coordinative patterns were analyzed using RMS differences and Cross-Correlation Coefficients. RMS differences in intralimb coupling were evaluated over stance and swing phases. A significant main effect of RMS differences for thigh-shank coupling during the stance phase in the loaded side was observed (F (2, 46) = 10.852, P < .001). Post-hoc testing revealed that thigh-shank coupling during condition 1 was less than conditions 2 and 3 (P = .01, P = .001; Figure 4-3). Also, on the unloaded side, a main effect of RMS in thigh-shank was detected (F (2, 46) = 5.378, P = .008). RMS changes during condition 1 were decreased compared to conditions 2 and 3 (P = .033 and P = .041, respectively; Figure 4-3). However, no significant RMS difference was observed for thigh-shank coupling during swing phase (P > .05).

For shank-foot coupling, significant differences in RMS were observed on the loaded side during both stance (F (2, 46) = 9.220, P < .001) and swing phases (F (2, 46) = 5.695, P = .006). RMS decreased during condition 1 compared to conditions 2 and 3 (P = .02 and P = .004, respectively, Figure 4-4). Also, during the swing phase, the RMS difference during condition 1 in the loaded side was less than condition 3 (P = .027,

Figure 4-4). However, no significant differences of RMS in shank-foot coupling were detected for the unloaded side (P> .05). Table 4-2 illustrates the loading conditions compared to each other for RMS difference. Moreover, no cross-correlation coefficient effects for thigh-shank and shank-foot coupling for either limb were displayed. All CCC values for intralimb coupling were close to 1.

Interlimb Coordination

Interlimb coordination was examined via thigh-thigh, shank-shank and foot-foot couplings. A significant main effect on RMS difference was observed in thigh-thigh coupling (F (2, 46) = 5.276, P = .009). RMS differences during condition 1 in thigh-thigh coupling were significantly less than condition 2 (P = .022; Figure 4-5). There were no other significant differences between conditions (P >.05). No effect on RMS changes in shank-shank or foot-foot was observed. Also Cross-Correlation Coefficient in thigh-thigh varied across the conditions (F (2, 46) = 4.651, P = .014). CCC for thigh-thigh coupling during condition 1 were significantly increased compared to condition 3 (P = .039; Figure 4-5). There was no difference for CCC in condition 1 versus 2 or condition 1 versus 3 (P >.05). No effect on Cross Correlation Coefficient in shank-shank and foot-foot coupling was displayed (P> .05). Table 4-3 illustrates the loading conditions compared to each other for RMS difference and CCC in interlimb couplings.

Table 4-1. Comparisons of the loading conditions for gait kinematics.

Kinematic variables	Condition 1 v. 2	Condition 1 v. 3	Condition 2 v. 3
Stride Length	↔	↔	↓
Cadence	↔	↔	↑
Step Width	↓	↓	↔
Swing/Stance Ratio-Loaded	↑	↑	↔↔
Swing/Stance Ratio-Unloaded	↓	↓	↔

Note. ↔ = no difference, ↑ = increase, ↓ = decrease

Table 4-2.Comparisons of RMS difference for intralimb couplings.

RMS Difference		Condition 1 v. 2	Condition 1 v. 3	Condition 2 v. 3
Loaded	**Thigh-shank**			
	stance	↓	↓	↔
	swing	↔	↔	↔
	Shank-Foot			
	stance	↓	↓	↔
	swing	↔	↓	↔
Unloaded	**Thigh-Shank**			
	Stance	↓	↓	↔
	Swing	↔	↔	↔
	Shank-foot			
	Stance	↔	↔	↔
	Swing	↔	↔	↔

Note. ↔ = no difference, ↑ = increase, ↓ = decrease

Table 4-3. Comparisons of RMS difference and CCC for interlimb couplings.

RMS difference	Condition 1 v. 2	Condition 1 v. 3	Condition 2 v. 3
Thigh-Thigh	↑	↔	↔
Shank-Shank	↔	↔	↔
Foot-Foot	↔	↔	↔
CCC			
Thigh-Thigh	↔	↓	↔
Shank-Shank	↔	↔	↔
Foot-Foot	↔	↔	↔

Note. ↔ = no difference, ↑ = increase, ↓ = decrease

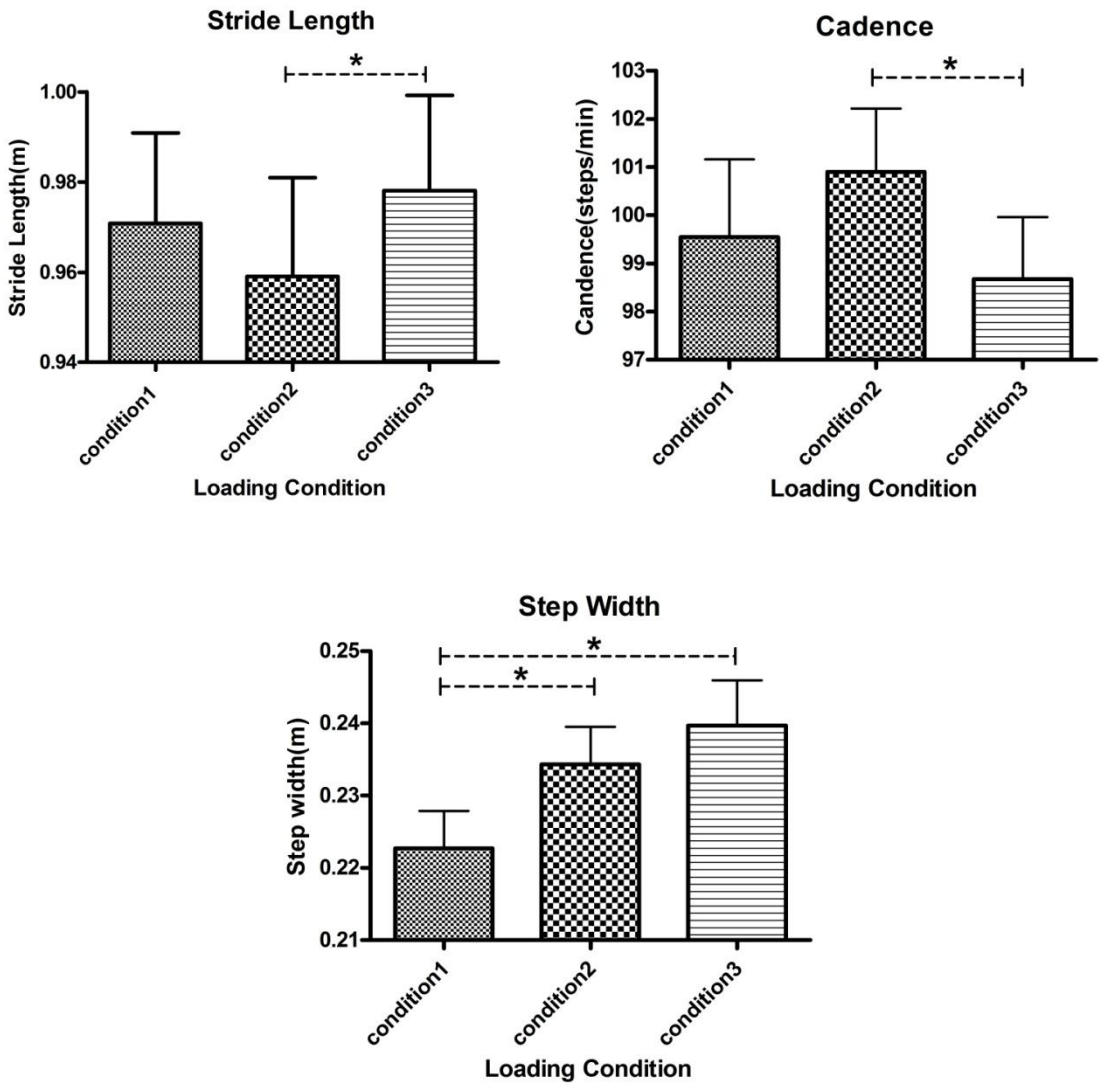


Figure 4-1. Effect of three different loading conditions (*Figure 3-1)* on stride length, cadence, and step width. *P< .05.

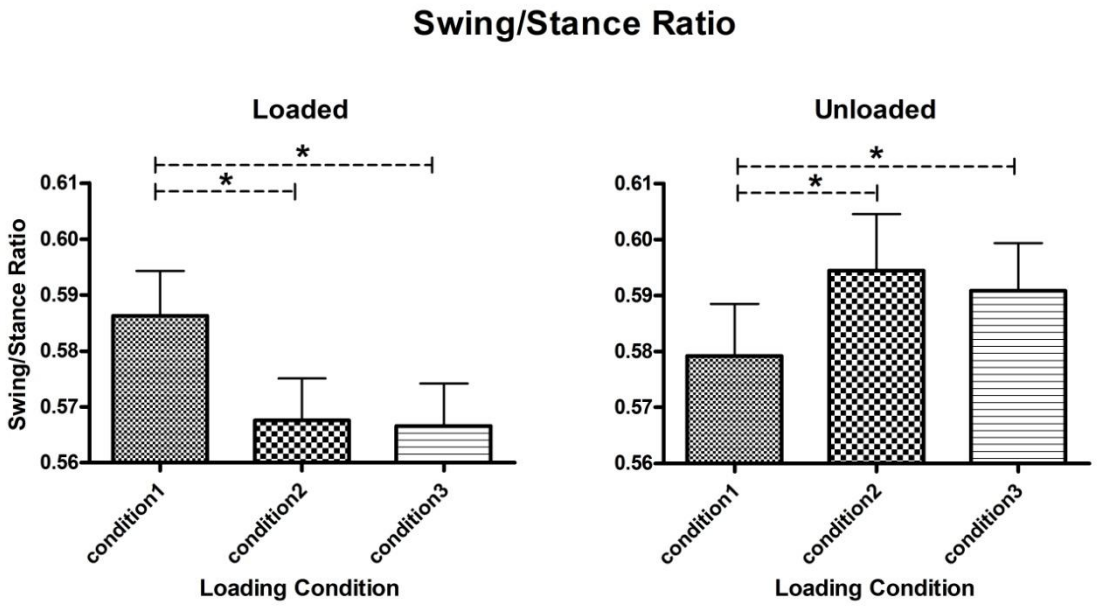


Figure 4-2. Effect of three different loading conditions (*Figure 3-1)* on swing/stance ratio. *P< .05.

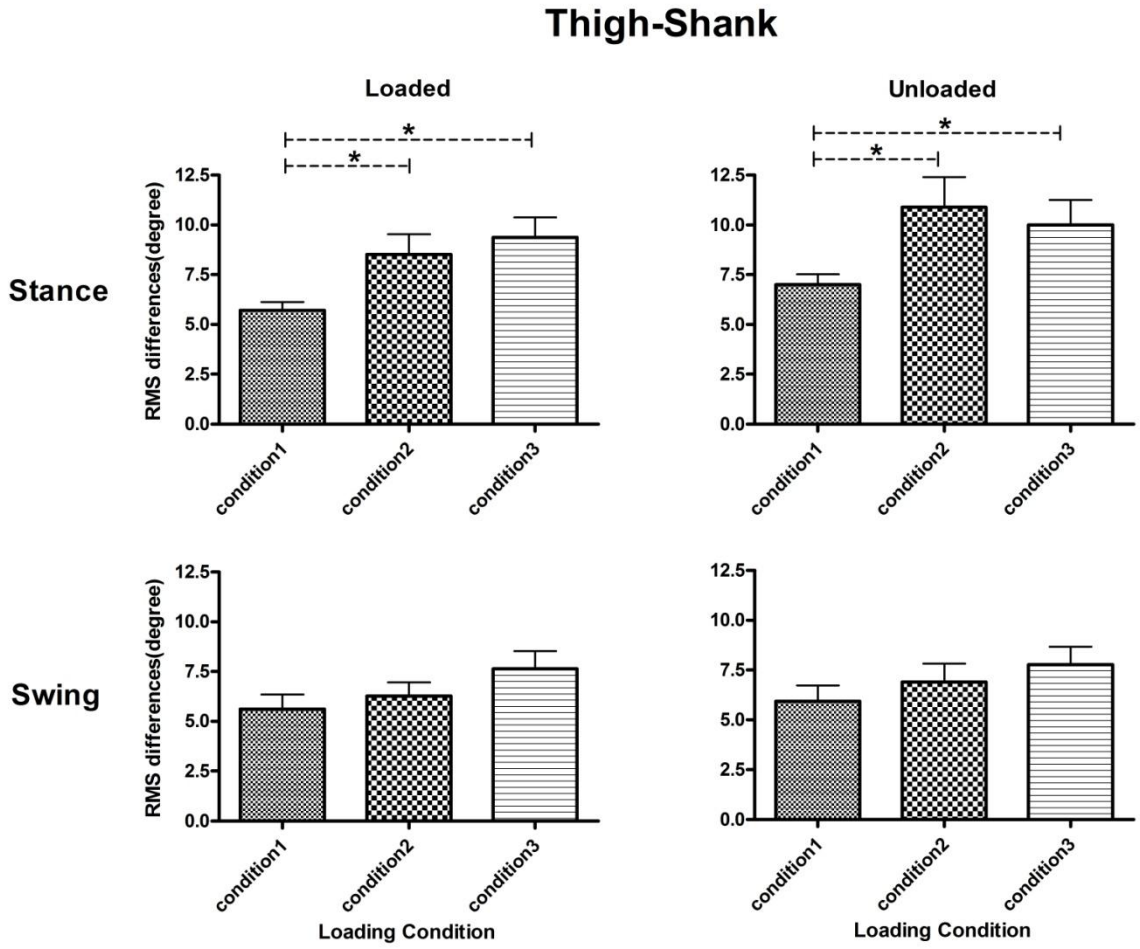


Figure 4-3. RMS difference in thigh-shank coupling during both stance and swing phase for the loaded and unloaded sides. *P<. 05.

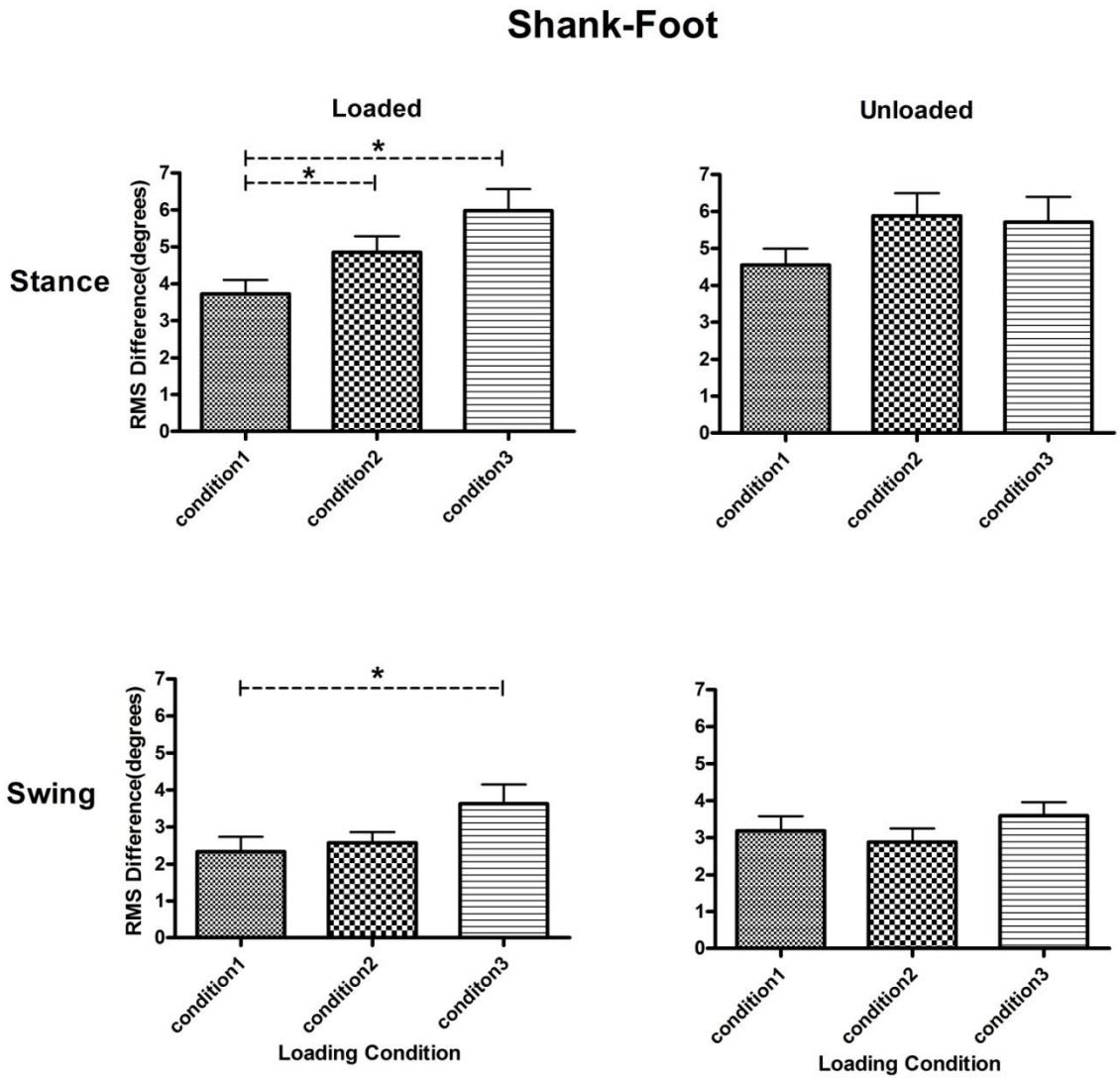


Figure 4-4. RMS difference in shank-foot coupling during both stance and swing phase for the loaded and unloaded sides. *P<. 05.

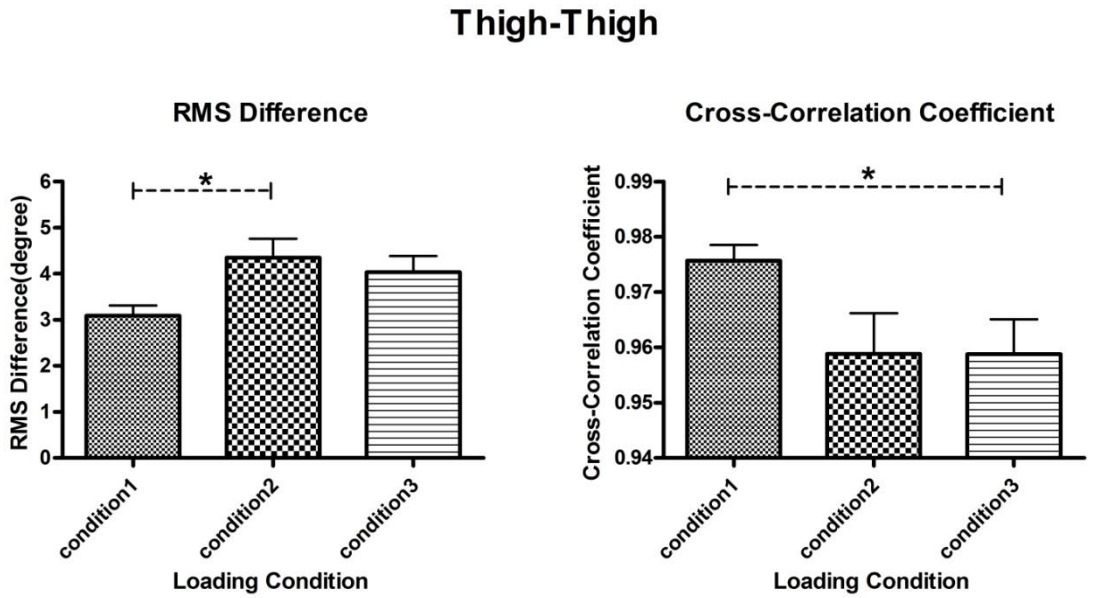


Figure 4-5. RMS difference and Cross-correlation coefficient for thigh-thigh coupling over a gait cycle. *P<. 05.

CHAPTER 5
DISCUSSION

Kinematic Gait Parameters: Hypothesis 1

The aim of this study was to investigate the effect of different loading conditions on temporal-spatial gait patterns during treadmill walking for healthy university students. Our first hypothesis was supported or partially supported for three of the five tests performed related to gait kinematics (Table 4-1). More specifically, Gait pattern was improved when carrying two messenger bags (one on each shoulder) compared to a messenger bag on one shoulder (hanging vertically or draped across the body). Several previous studies have identified the influence of loading on gait kinematics (Crosbie et al., 1994; Pascoe et al., 1997; Cottalorda et al., 2003; Connolly et al., 2008; Birrel & Haslam, 2009). Stride length and cadence in normal gait are approximately 1.41m and 113steps/min in adult participants, respectively (Perry, 1992). Our values regarding stride length and cadence were less than those of normal walking; presumably because participants in the current study walked on a treadmill. On a treadmill, individuals show decreased stride length and faster cadence due to short treadmill belts (Cottalorda et al., 2003). Accordingly, we observed lower cadence values across conditions during treadmill walking than those of normal gait. In addition, the individuals tested here selected a slower speed (0.8m/s) than normal walking velocity as their preferred walking speed.

Crosbie and colleagues examined the effect of unilateral load carriage at 10% and 20% of body weight for twenty college students. They reported a decrease in step length and an increase in cadence in response to unilateral carriage while walking barefoot on a flat walkway (Crosbie et al., 1994). Pascoe et al. (1997) found that stride

length was decreased and cadence was increased when carrying a one strap bag, two-strap bag, or a one-strap athletic bag compared to no bag. Additionally, these authors did *not find a significant difference between unilateral and bilateral load carriage*. Our results, in part, support this tendency regarding stride length and cadence and will be discussed shortly. However, in the current study, the no load condition for gait kinematics was not measured and thus was not available for comparison. In a more recent study, Cottalorda et al. (2003) evaluated the effect of different methods of backpack carrying (unilateral and bilateral carriage) while walking on a treadmill on gait kinematics in children. Again, a decrease in stride length and an increase in cadence when carrying a backpack on both shoulders compared to no bag was found, but *no difference between one strap and two straps* was observed and only a single one strap condition was evaluated. Furthermore, Connolly et al. (2008) studied thirty-two children under two different load conditions while walking on an electronic walkway (GAITRite system) and reported a decrease in stride length when carrying the backpack loaded with 15% of body weight on one shoulder compared to those without a backpack. However, they also reported *no difference in stride length between two experimental conditions (*carrying a backpack on one shoulder and two shoulders*)*. Our data were consistent with these previous studies and did not show significant differences in stride length and cadence between unilateral load carriage and bilateral load carriage. Therefore, regardless of whether participants walked overground or on a treadmill, stride length and cadence were not sensitive to symmetric and asymmetric loading. In addition, it is possible that the mechanics of walking on a treadmill were externally dictated. The velocity of the treadmill was fixed and may not have matched the subject's

preferred walking speed in each condition. Thus, these parameters (stride length and cadence) were not influenced by different methods of load carriage while walking on a treadmill. However, we also evaluated the effect of different unilateral load carriage strategies (conditions 2 and 3). When carrying the messenger bag on one shoulder hanging vertically (condition 2), stride length was decreased compared to when the participant carried a messenger bag on one shoulder with the bag draped across the trunk to the contralateral hip (condition 3). This result was unexpected, could infer potential benefits related to condition 3, and warrants further research. Changes in stride length and cadence during walking are associated with the sagittal plane and may contribute to gait stability. It has been shown that shorter stride length decreases anterior-posterior stability (McAndrew & Dingwell, 2012). Moreover, faster cadence may indicate an adaptive strategy to maintain dynamic balance by selecting a proper cadence in response to external constraints (Arif et al., 2002; Rogers et al., 2008). None of our participants had difficulty completing the walking tasks even though altered stride length and cadence have been related to risk of falling (Maki, 1997). The shorter stride length and faster cadence observed during condition 2 compared to condition 3 indicate that this strategy used for unilateral load carriage could negatively alter walking kinematics and increase risk.

In addition, we found altered gait parameters in the frontal plane. Step width during asymmetrical load carriage (conditions 2 and 3) was increased compared to symmetrical load carriage (condition 1). In general, step width in normal walking ranges from 7.1 to 9.1cm (Murray et al., 1964). In the current study, all measurements of step widths across the conditions were greater than those of normal gait. Several previous

researchers have examined different methods of carrying loads in terms of step width. Crosbie et al. (1994) reported a significant decrease in step width for unilateral loading conditions (10% and 20% of body weight) compared to a no load condition for male college students and a non-significant trend towards an increase in step width for female students. Connolly et al. (2008) observed decreased step widths while carrying a 10kg backpack on one shoulder compared to without a backpack for children. Also, a more recent study showed a decrease in step width while carrying loads (10% and 20% of body weight) in one hand compared to no load for healthy adults (Zhang et al., 2009). As stated previously, in our current study, we did not evaluate the no load condition. Thus, in this respect, our data could not be directly compared to previous research. Despite this inconvenience, important inferences can be made. In general, unilateral loads shift the center of mass towards the loaded side. Thus, increased step width during asymmetrical loading conditions may indicate an increase in the base of support boundaries to maintain lateral stability during unilateral carriage (Crosbie et al., 1994; Connolly et al., 2008). Therefore, these alterations in step width may show a compensatory adjustment of shifted COM in response to asymmetric load carriage. To preserve dynamic balance, our participants may have needed the increased step width to compensate for the laterally displaced COM.

We also found an effect of asymmetrical loading on stance and swing time over a gait cycle. Normal gait consists of 62% stance phase and 38% swing phase, indicating 0.61 swing /stance ratio (Kirtley, 2008). In the current study, Swing/stance ratios were measured in three different loading conditions for both loaded and unloaded sides and were slightly below 0.61 for each. A higher swing/stance ratio indicates more swing time

and less stance time over a gait cycle and decreased stability (Berman et al., 1987; Roth et al., 1997; Wu et al., 2000). On the loaded side, the swing/stance ratios for asymmetrical loading (conditions 2 and 3) were increased compared to symmetrical loading which indicates more stance time and less swing time over the gait cycle. For the unloaded side, we found an opposite trend indicating that swing/stance ratios in condition 2 and 3 were decreased compared to condition 1. Asymmetrical load carriage affected both legs in a complementary manner because during one leg's stance phase the other leg has swing phase over a gait cycle. The effect of loads on swing/stance ratio or stance and swing time has not been investigated in other studies. However, a similar effect on swing and stance time has been seen for obese people (Błaszczyk et al., 2011). These authors found 5% shorter swing and 3% longer stance for obese individuals compared to lean participants. In the current study, the average effective BMI for participants carrying a load of 10% of body weight was increased to 25.3. Thus they were classified as overweight (BMI of 25-29.9) but not obese (BMI > 30). We found similar changes on the loaded side which indicates more stance time to possibly avoid losing dynamic stability. Moreover, complementary changes on the unloaded side may be an additional adaptive strategy to maintain dynamic balance during gait. The changes in both legs indicate an asymmetric gait pattern in terms of swing/stance ratio. This change in symmetry relative to swing/stance ratio has been reported in hemiplegic gait (Roth et al., 1997). Also, this temporal asymmetry was suggested in patients with chronic stroke (Patterson et al., 2008). Thus, asymmetrical loading appears to cause inverse swing/stance ratio changes in the loaded and unloaded sides. Although, these alterations in symmetry assist to preserve dynamic balance, ultimately these changes

are less stable and this may lead to abnormal walking patterns in response to the asymmetric load on one shoulder.

To conclude, gait patterns were altered by different loading conditions in the current study (primarily in the frontal plane). Alterations in temporospatial parameters during different load conditions may provide insight into adaptive strategies in response to external constraints. These alterations may increase the risk for injury in some cases. Most of the changes in gait patterns were detected in condition 2 compared to the other conditions. Based on our findings, altered gait patterns observed while carrying a messenger bag on one shoulder hanging vertically to the hip were needed to compensate for decreased stability during gait.

Limb Coordination: Hypothesis 2

The present study of intralimb and interlimb coordination using CRP aimed to evaluate coordinative lower extremity mechanisms in response to different loading conditions during treadmill walking. Hypothesis 2 was supported by three out of four dependent measures for stance and partially supported for two of four related to swing in terms of RMS for intralimb coupling (Table 4-2). Also, hypothesis 2 was supported by two out of six dependent measures related to interlimb couplings in terms of RMS and CCC measures (Table 4-3). RMS difference and CCC were used to quantify differences in intralimb and interlimb coordination for the three different loading conditions and indicated that coordination was altered during unilateral load carriage. Specifically, two *average* CRP curves in the baseline condition and each carriage condition were compared to evaluate coordinative patterns for the three different loading conditions using RMS and Cross-correlation coefficient. While the CCC measure indicates changes in the spatio-temporal evolution of CRP patterns, RMS measures show

information about the magnitude differences in relative phase between the patterns (Haddad et al., 2004).

For thigh-shank coupling, RMS difference in both loaded and unloaded sides during stance phase were increased in condition 2 and condition 3 compared to condition 1. We found no difference in RMS in the swing phase for thigh-shank coupling. Although, we utilized different experimental conditions (bilateral and unilateral load carriages) than those used by Haddad and colleagues (unilateral load carriage) we can gain insight into the effect of unilateral load carriage. Specifically, similar tendencies for RMS differences between no load and unilateral leg loads conditions have been detected in the previous research (Haddad et al., 2004). They utilized six unilateral leg loads (no load, 0.9, 1.8, 2.7, 3.6, and 4.5 kg) during treadmill walking and observed an increase in RMS changes (4-9°) over both stance and swing phase as the leg load was increased for the loaded limb in thigh-shank coupling with no effect on the unloaded side. We also observed RMS changes on the unloaded side in stance and no difference in RMS in swing phase in thigh-shank couplings. In part, their findings are in contrast with our results due to different methods of carrying loads. However, RMS difference only represents information on the magnitude of difference between two curves in different loading conditions but does not show which curve is more out-of or in phase. Complete CRP curves provide information regarding how in phase or out of phase two segments are during the *entire* stance phase. In phase coupling indicates that two segments of the body are temporally aligned regarding three components: angular displacement, angular velocity and direction of two segments movement; non-alignment is indicative of out-of phase coupling. This additional information provides

insight into the coordination patterns utilized in the different conditions tested here (Figures 5-1 to 5-5). For example, the CRP pattern in condition 2 (Figure 5-1) was more out-of phase than normal walking (baseline) in early stance phase (0-20%) for thigh-shank coupling in the loaded side. Less out-of phase thigh-shank coupling was observed on the loaded side for asymmetrical loads (conditions 2 and 3) compared to the symmetrical load (condition 1).The majority of the RMS difference in condition 2 for thigh-shank coupling on the loaded side was in mid to late stance (40-90%). Also, RMS changes in condition 3 compared to condition 1 resulted from less out-of phase coupling during stance (30-60% and 80-100%).These RMS changes indicate restricted thigh-shank coupling in the loaded side during the unilateral load carriage on one shoulder. We also found similar RMS modification in thigh-shank coupling on the unloaded side. For the two asymmetric loading conditions, the CRP curves for condition 2 and condition 3 (Figure 5-2) were less out-of phase for thigh-shank coupling compared to condition 1. Major changes in condition 2 compared to condition 1 appeared in mid stance (40-60%). Also, RMS changes in condition 3 compared to condition 1 appeared in mid to late stance (30-100%). Less out-of phase thigh-shank patterns have been seen in hemiparetic gait (Hutin et al., 2010). Moreover, Hutin and colleagues applied an orthotic knee constraint in healthy participants, which produced a reduction in RMS during swing and stance phases. The loss of thigh-shank coupling during stance phase indicates one of the altered coordination patterns that may relate to restricted knee flexion during gait. Therefore, asymmetrical loading may contribute to less out-of phase movement, altering knee flexion during stance phase in both the loaded and unloaded side compared to symmetrical loading.

For shank-foot coupling, we found significant RMS differences only on the loaded side during both swing and stance phase. These RMS differences also represent partial information. Further graphical analysis (Figure 5-3) revealed that the RMS differences in condition 2 and 3 result from more out-of phase coupling compared to condition 1. The CRP curves in conditions 2 and 3 during late stance phase (80-100%) show more out-of phase movement compared to condition 1. These two asymmetrical loading conditions contributed to intensified out of phase shank-foot coupling on the loaded side during late stance (Figure 5-3). A significant RMS change in condition 3 on the loaded side resulting from changes during early and mid-late swing phase (0-20% and 60-90%) was also observed compared condition 1 which indicates a more out-of phase pattern in condition 3 (Figure 5-4). Regarding intralimb coupling, we also detected CCC changes in response to difference loading conditions. Our findings support previous research (Haddad et al., 2004) that displayed invariance in CCC (all values were greater than 0.99) for the intralimb couplings while carrying a unilaterally applied leg load. Also, these authors showed altered RMS for loaded limb coupling in response to increases in leg load. However, we found an RMS difference in thigh-shank coupling on the unloaded side as well as the loaded side. Also, Haddad et al. (2004) indicated only RMS changes but not where these RMS changes resulted from. Another interesting finding here is that CRP curves in thigh-shank and shank-foot display completely different adaptations in response to asymmetric loading. Thigh-shank couplings on both sides during stance were *less* out-of phase, while shank-foot coupling showed a *more* out-of phase pattern on the loaded side during both swing and stance with an asymmetrical load compared to symmetric loading. Thus, it is possible that increased

out-of phase shank-foot coupling may be related to increased ankle stiffness during asymmetrical load carriage in the loaded limb which could result in restricted knee movement.

The effect of asymmetric load carriage was also observed for interlimb coordination. As mentioned before, RMS changes in thigh-thigh coupling while carrying the asymmetric load in condition 2 were observed. More specifically, these changes resulted from exaggerated asymmetry in thigh-thigh coupling during the gait cycle (Figure 5-5). Also, the CCC value in thigh-thigh coupling was decreased when carrying a messenger bag on one shoulder (condition 3), indicating a higher difference in coordination compared to carrying two messenger bags (one on each shoulder). Previous researchers reported an increase in RMS and a decrease in CCC regarding interlimb couplings in response to unilateral leg loads during treadmill walking (Haddad et al., 2004). They observed an increase in RMS and a decrease in CCC (.95-0.7) for thigh-thigh, shank-shank and foot-foot couplings, as leg load was increased. However, in the current study, increased asymmetry in interlimb coordination was found only for thigh-thigh coupling in terms of RMS and CCC measures. In general, smoothness and symmetry in gait are regarded as ‘normal walking’. Asymmetry during gait has been observed in previous pathological gait research as stroke, Parkinson’s, hemiplegia and cerebral palsy (Roth et al., 1997; Patterson et al., 2008; Johnsen et al., 2009; Meyns et al., 2012). In these previous studies, researchers have focused on clinical treatments to improve the asymmetry in gait for these patients. Asymmetry during gait, as observed here, may be considered potentially injurious with symptoms beyond the capacity of locomotor system.

The results of the current study suggest a variety of adaptations in intralimb and interlimb coordination in response to symmetric and asymmetric load carriage. We expected that altered coordinative patterns in the lower extremities would be displayed during asymmetrical load carriage. As indicated before, changes in intralimb and interlimb coordination were observed for the asymmetrical loading conditions compared to symmetric loading condition. However, the two different asymmetrical conditions did not show any difference in interlimb and intralimb couplings. We observed abnormal patterns in thigh-shank, shank-foot, and thigh-thigh coordination. Based on our findings, we recommend people avoid carrying a messenger bag on one shoulder (condition 2 or 3) in order to decrease abnormal limb coordination in daily activities. These alterations may provide researchers with preliminary knowledge concerning diverse gait adaptations caused by external constraints. The adaptations in limb coordination during asymmetrical load carriage should be further investigated, as they may be indicative of the possibility for acute and/or chronic joint injury and pain. Furthermore, asymmetric load carriage may have potential to alter coordinative mechanisms in the lower extremities which may be associated with the risk of falling in taxing circumstances (e.g. slope, stairs, wet floors, etc.) Our results provide a picture of adaptive mechanisms of locomotor systems under various constraints, which may enable us to educate at-risk individuals (e.g., the elderly and children) on the dangers of abnormal gait patterns.

Limitation of Study

In our study, arm movement was constrained across the previously referenced conditions because clear and accurate collection of kinematic data in the laboratory environment required participants to cross their arms. However, arm movements are indispensable parts of human locomotion that play an important role for contralateral

limb movements (Eke-Okoro et al., 1997). In addition, as alluded to previously, we evaluated individuals walking at a fixed velocity on a treadmill. Thus, they could not alter their walking velocity. However, changing walking speed in response to different loading conditions could be an important adjustment to preserve dynamic balance during these conditions. Subsequent work could involve calculating a preferred walking pace in each experimental condition. Finally, the differences between overground and treadmill walking should be considered. On a treadmill, altered gait patterns have been observed because of short treadmill belts (Cottalorda et al., 2003). Thus, it is possible that different coordinative patterns during overground walking may exist. Further, treadmill walking may have limited application to real world conditions when people carry loads for work or recreation.

Thigh-shank in loaded side

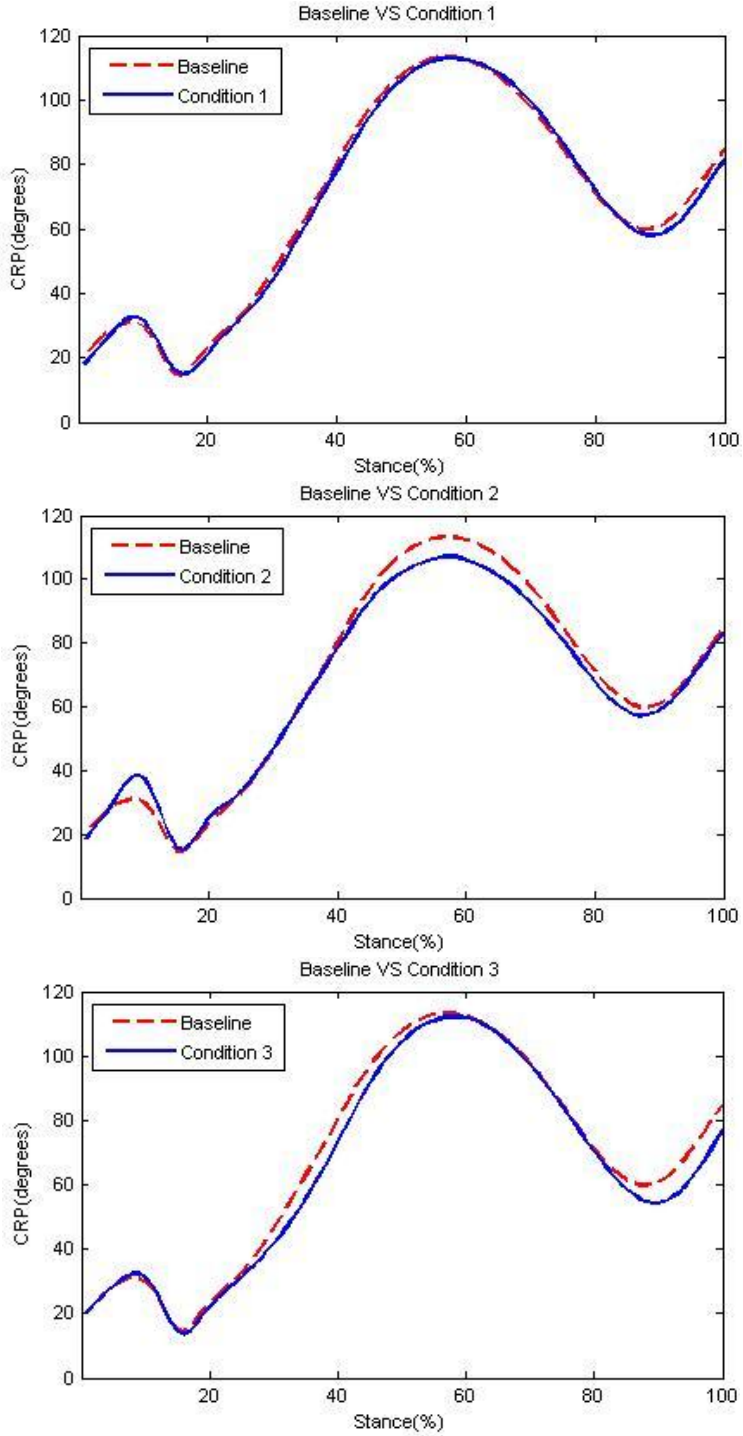


Figure 5-1. Mean CRP curves in thigh-shank on the loaded side during stance phase for baseline (no load) and each experimental condition (n=24).

Thigh-shank in unloaded side

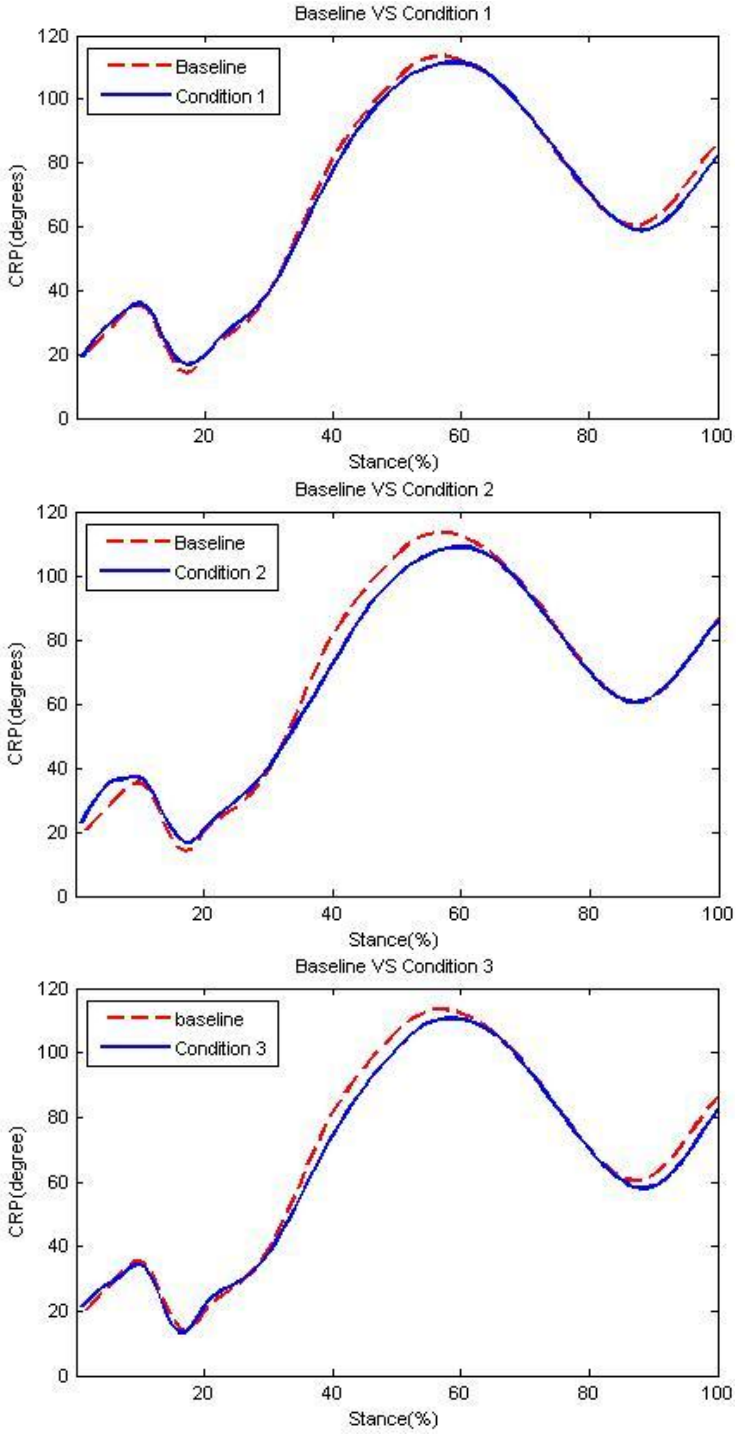


Figure 5-2. Mean CRP curves in thigh-shank on the unloaded side during stance phase for baseline (no load) and each experimental condition (n=24).

Shank-foot in loaded side

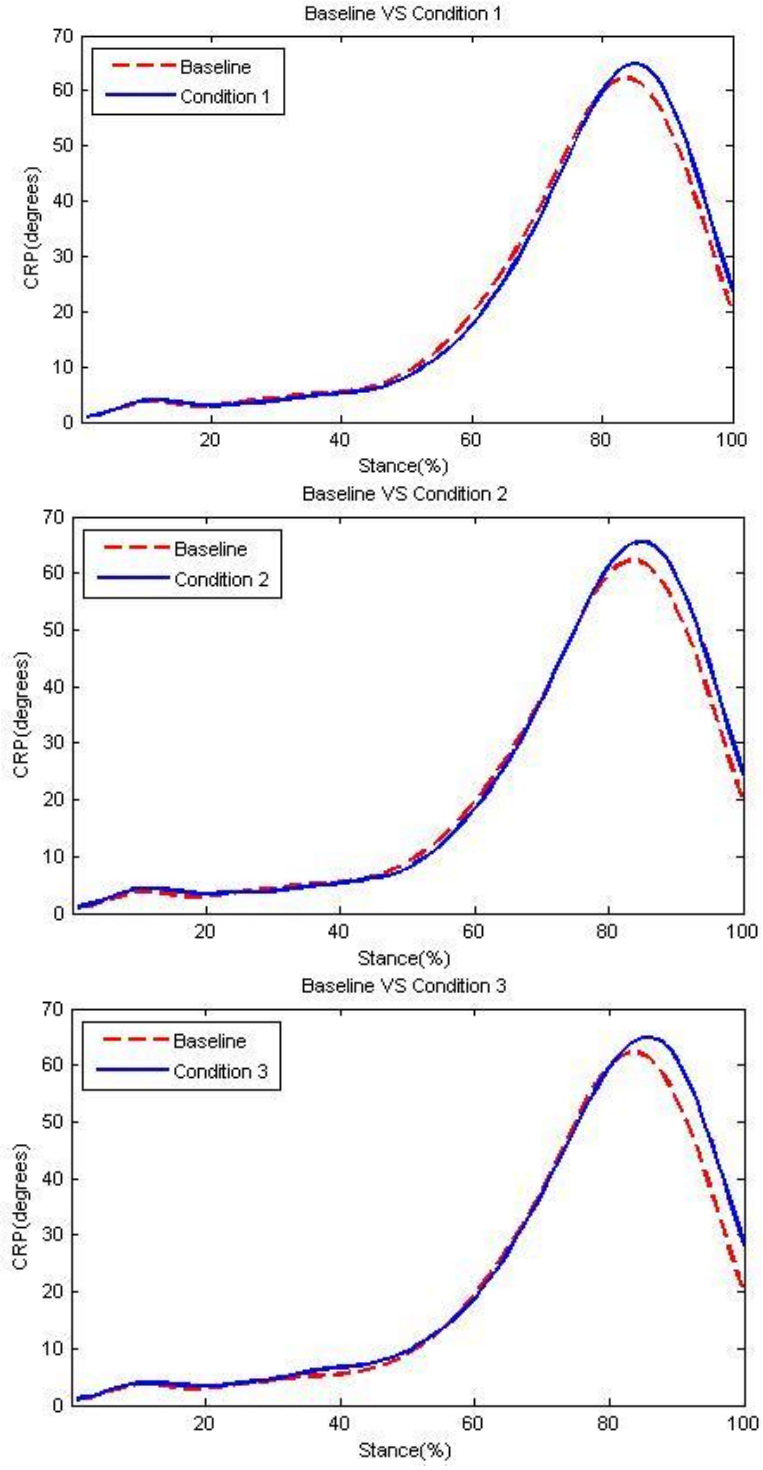


Figure 5-3. Mean CRP curves in shank-foot on the loaded side during stance phase for baseline (no load) and each experimental condition (n=24).

Shank-foot in loaded side

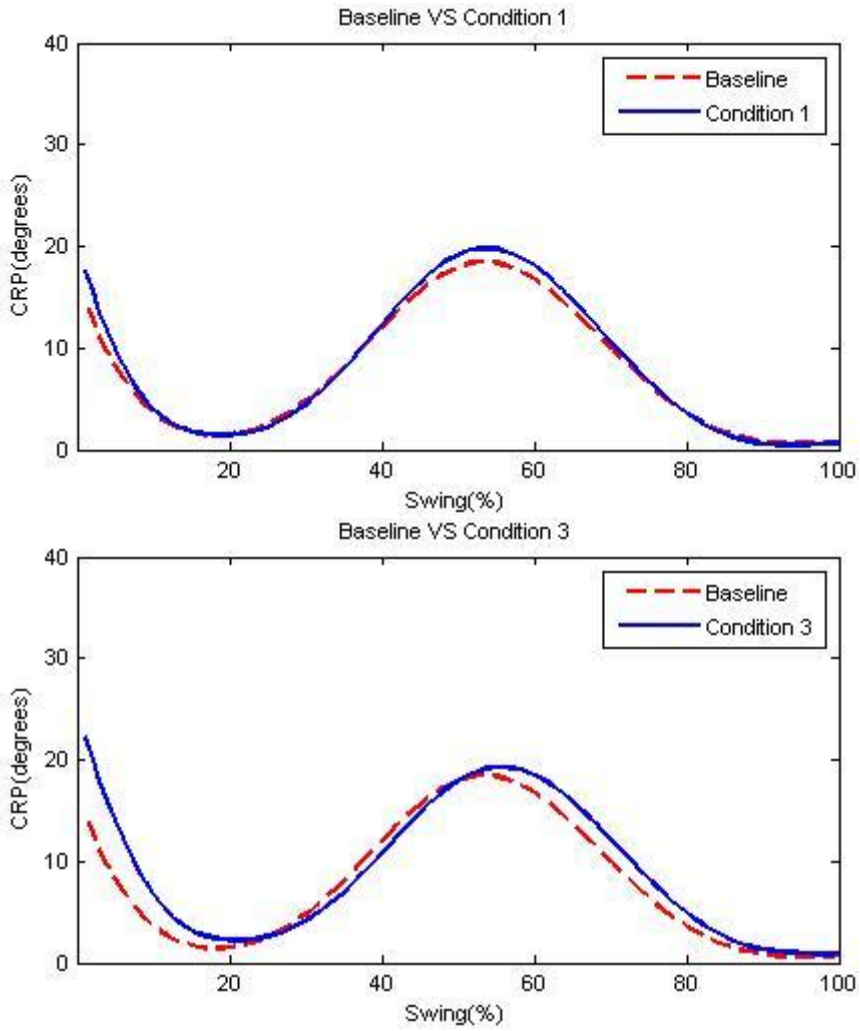


Figure 5-4. Mean CRP curves in shank-foot on the loaded side during swing phase for baseline (no load) vs condition 1 and 3 (n=24).

Thigh-thigh

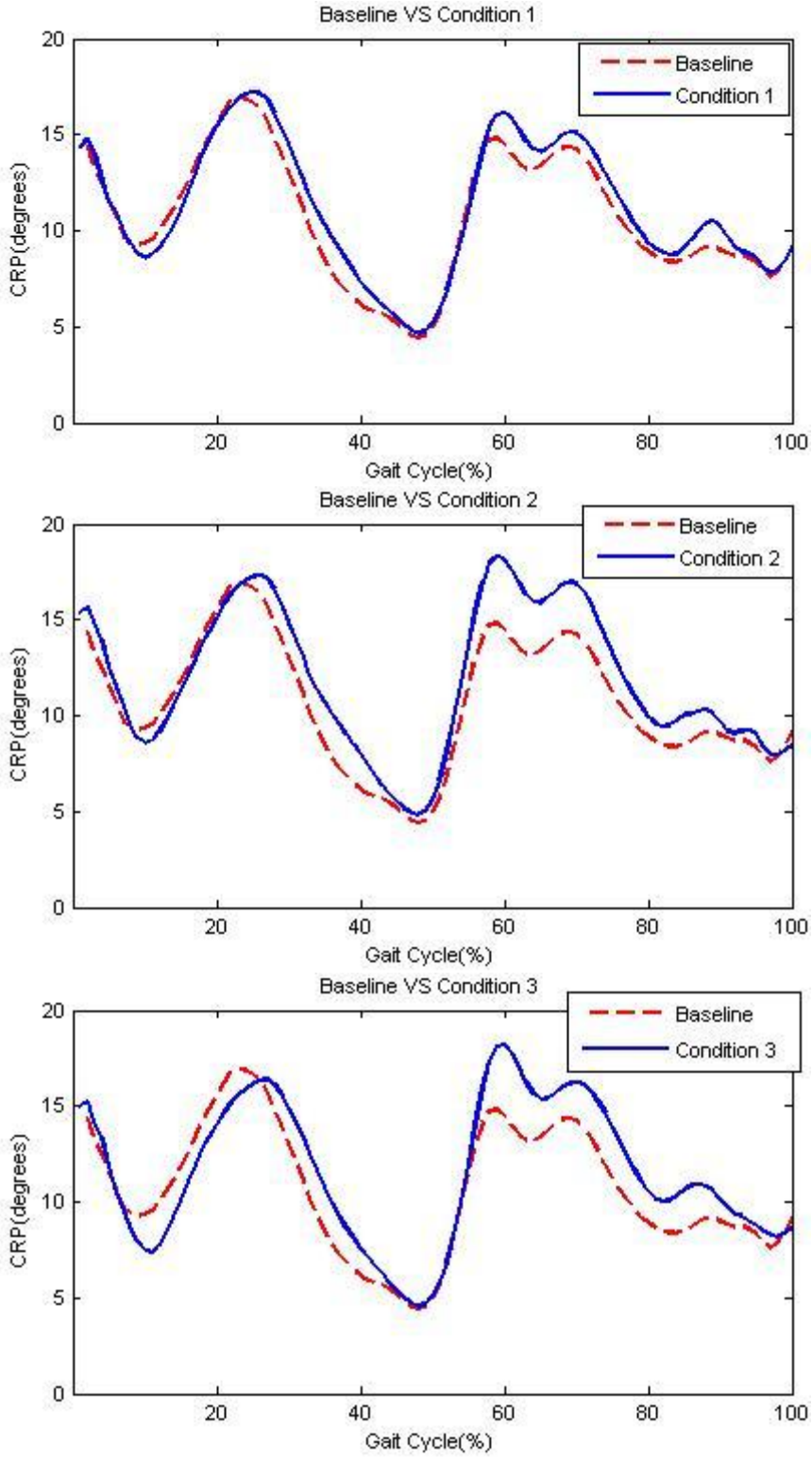


Figure 5-5. Mean CRP curves in thigh-thigh coupling over a gait cycle for baseline (no load) and each condition (n=24).

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