Biomechanical Analysis of Asymmetrical Lateral Loading on Gait: A Pilot Study

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Abstract

A significant number of studies have linked low back pain and load carriage. Many of these studies, however, have focused on posterior or anterior load carriage, which is different from the standard load carriage for populations such as law enforcement officers and construction workers, to name a few. Thus, this pilot study focuses on the biomechanical analysis of lateral duty belt load carriage in order to investigate whether there exists a weight difference threshold between lightly loaded and unloaded limbs for observing compensatory gait adjustments. One subject participated in four different load conditions. Biomechanical variables of interest included contralateral pelvic drop, hip internal rotation and external rotation, trunk excursion in the frontal plane, and peak ground reaction forces for both the weighted and unweighted limbs.

I. Introduction

According to a 2011 article in the Journal of Criminal Justice Research, 86% of the studied police officers reported having back pain, with 61% of the officers attributing their pain to the police belt and vest, in addition to the seat in the police car [1]. For security personnel, duty belts are an essential part of the uniform, as they act as a viable mechanism for transporting equipment. On average, with all of the equipment, officers carry between 10 and 30 pounds. However, there is no standard way to place items on a duty belt, and each person utilizes the belt in a way that matches their individual needs. For example, officers might place pepper spray on the side of their dominant hand so that it is able to be easily accessed when needed. This lack of standardization can lead to asymmetrically weighted belts and may negatively impact officers' performance and safety. Given that there exists an established link between load carriage and low back pain [1, 2], it is important to determine how the load carried by police affects performance of tasks typically encountered. This information could significantly aid future researchers in the development of safer and healthier load carriage designs for police.

When loads are asymmetrical, gait alterations can be observed. Specifically, changes to joint kinematic patterns, such as an increase in hip flexor and knee extensor activity in early swing followed by an increase in hip extensors and knee flexor activity in late swing have been seen after a 2kg mass was added to the left lower leg [3]. In addition, challenges to upper body stability and increases in vertical ground reaction forces tended to be strongly associated with the increasing weight borne on the executing limb [4,5]. Furthermore, level and uphill front-loaded weight carriage has been correlated with a decreased stride and step length, in addition to increased cadence, particularly during level and uphill walking [6]. The same study also compared front-loaded walking with level and uphill unloaded and back-loaded walking; after analysis, it was ultimately observed that postural control, as indicated by increased erector spinae EMG activity and changes in spatiotemporal parameters of gait (stride length and rate), was compromised particularly during level and uphill front-loaded walking [6].

However, much of the literature has discussed loading in detail with regards to unloaded, front-loaded, and back-loaded walking, but very little has been studied about lateral-loaded walking, which is a gap that this study aims to fill. Additionally, although there is a wealth of literature exploring the impact of added load (backpacks in particular) on human performance [7, 8, 9], only a few studies have examined the effect of asymmetrical, lateral loading on tasks commonly encountered in policing [10, 11]. While much can be learned from backpack studies, the results cannot be generalized to police who typically carry smaller loads that are positioned anteriorly around the hips. Therefore, Ramstrand et al. observed a significant difference between the standard belt condition and

of the hip joint [11]. Additionally, Dempsey et al. sought to study duty belts with stab-resistant body armor in order to measure dynamic balance and physiological cost during 5 mobility tasks [10]. Ultimately, this study observed that participants were off-balance longer, slower to complete the acceleration, grapple and mobility tasks, completed fewer chin-ups, and had greater physiological cost [10]. Most research, though, involves the analysis of military-based tasks using heavier and more specialized military equipment, finding that it adversely affects mobility, strength, speed, balance and physiological strain [12, 13].

Yet, from the limited amount of research found, in response to wearing a duty belt, previous research has found correlations with restricted mobility and physiological costs, lower back pain, altered gait kinematics, and spatiotemporal changes [1, 10, 11, 14, 15]. However, none of these studies have focused solely on laterally loading one hip and studying the resulting effects on kinematic gait variables and ground reaction forces. Therefore, the purpose of this study is to quantify the symmetry, or lack of, kinematic and kinetic variables between the lateral sides of the body in a healthy subject during a level walking task. The research is specifically aimed at determining if even relatively small asymmetries impact movement symmetry. Based on previous literature, it is hypothesized that in order to support the increase in weight when a load is added laterally to the human body, compensatory gait adjustments will be made in order to preserve balance. Specifically, it is thought that contralateral pelvic drop and trunk excursion in the frontal plane, hip internal and external rotation, and the peak ground reaction forces (GRFs) will be significantly different between the weighted and unweighted limb. Furthermore, it is thought that there will be greater observed compensatory adjustments that result from a larger weight difference between the two limbs.

II. Methodology

A. Participants

Healthy adults age 18 or greater were considered eligible to participate in the study. Individuals with recent musculoskeletal injuries were excluded. Subjects were also not paid to participate. The subject who volunteered for this pilot study was a college-age female who weighed 52.2 kg and had a height of 160 cm.

the unweighted condition in relation to internal and external rotation

B. Procedure

First, a static calibration trial will be collected. For the movement trials, the subject will perform 4 level walking tasks three times each over three force plates: one time with no weight as a control and three times with weight. More specifically, for the first trial, the participant walked across the three force plates while completely unweighted. For the second trial, the participant walked across the force plates with a duty belt and 4 lbs of additional weight placed on the belt at the right hip. For the third trial, the participant walked across the force plates with a duty belt and 2 lbs of additional weight on the belt at the right hip. For the fourth trial, 4 lbs were placed on the belt at the right hip and 2 lbs were placed at the left hip. In total, a weight of 6 lbs was placed on the subject, an amount that is slightly under the average weight for current police officers which poses no potential harm to untrained subjects. Gait speed was also controlled, and the participant walked with a velocity between 1.3 and 1.6 m/s for all trials. Walking velocity was monitored using timing gates (model) and was maintained within ± 10% of the pace that the participant self-selected. The participant remained shod for all static and movement trials. The loaded limb will be the one with the greatest amount of weight, and the unloaded limb is the one with the least, or no, weight added.

C. Data Collection and Processing

Biomechanical data were collected using a 12 camera Vicon MX3 (Vicon Motion Systems, Oxford, UK) motion analysis. 3D time-displacement data was collected from 24 infrared-reflecting hemispherical passive markers (Ø 15 mm), which were applied bilaterally on the head, torso, pelvis, lower extremities, and shoes of the participant (Fig. 1). Following a standard calibration trial, the anatomical markers were removed and the subject began the 4 movement trials. Ground reaction force data was collected using 3 force plates (model).

All kinematic and kinetic data were processed using Visual3D software (version 6; C-Motion, Inc, Germantown, MD). Raw marker-trajectory and force data were filtered at 6 and 50 Hz, respectively, using a fourth-order Butterworth

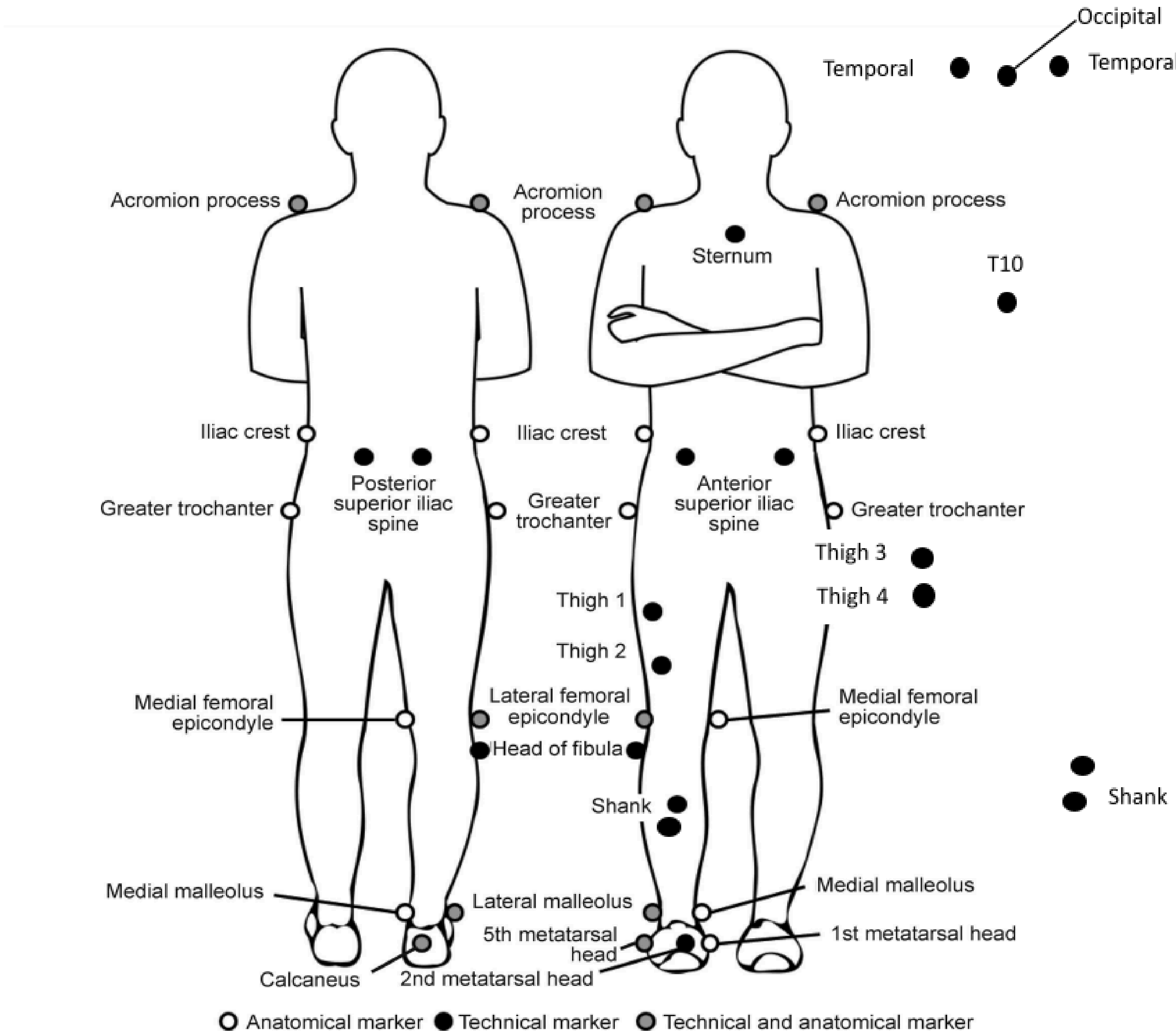


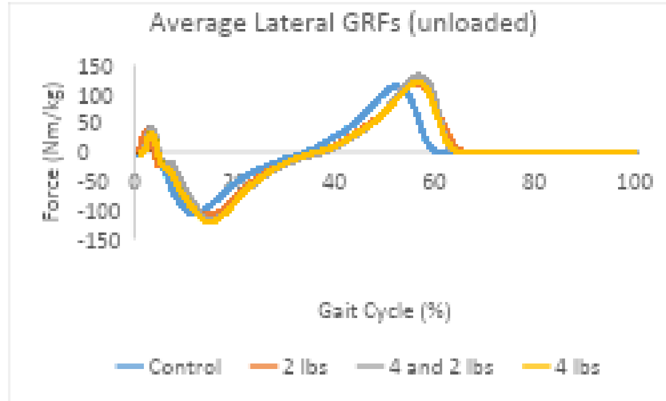
Figure 1. Placement of the anatomical and technical markers used to define and track segments, respectively: frontal, occipital, and temporal bones of the skull, acromion processes, sternum, T10 spinal segment, iliac crests, anterior- and posterior-superior iliac spines, greater trochanters, thigh cluster 1-4, lateral and medial femoral epicondyles, head of the fibula, shank cluster 1-4, lateral and medial malleoli, calcaneus, and first, second, and fifth metatarsal heads.

filter. All 3-dimensional joint angles were referenced as the distal segment relative to the proximal segment with the Cardan sequence of rotations using a Z-X-Y (mediolateral, anteroposterior, vertical) convention. Pelvic and trunk angles were defined as the respective segment relative to the laboratory. External joint moments were resolved into the distal coordinate system and were expressed as a percentage of body weight by height. Joint kinematics and moments were analyzed for the stance phase and normalized to 101 data points. To determine the stance phase, we identified initial contact and toe-off using a force threshold of 20 N.

A custom MATLAB program was used to graph the loaded and unloaded (right limb and left limb, respectively) hip, knee, and ankle angles for each movement trial condition. Microsoft Excel was used to extract the kinematic variables of interest for each condition. These included the peak angles for loaded and unloaded hip internal rotation and external rotation, in addition to the angular excursion (displacement between initial contact and the peak value) for contralateral pelvic drop and trunk excursion in the frontal plane. In addition, one-way ANOVA tests were performed for between-group statistical comparisons for all variables of interest. Significance was set at an alpha level of $P < 0.05$ and all statistical analysis was undertaken using Excel. After performing the ANOVA test, a Tukey post-hoc test was calculated to determine significant differences between specific loaded conditions. Lastly, the peak GRFs were also calculated for both the loaded and unloaded limbs.

III. Results

For all movement trials, significant differences were found between groups for pelvic drop excursion and trunk excursion in the frontal plane (Table 1). In regards to pelvic drop excursion (Fig. 2), significant differences were found when comparing the Control-2lbs, Control-4lbs,



2lbs-4lbs, 2lbs-4and2lbs, and 4lbs-4and2lbs trials ($P = 0.09*10^{-55}$). In regards to trunk excursion in the frontal plane, significant differences were found when comparing the Control-4lbs and the Control-

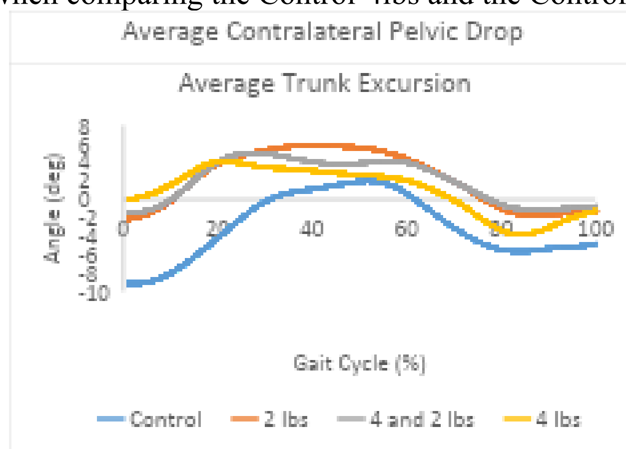
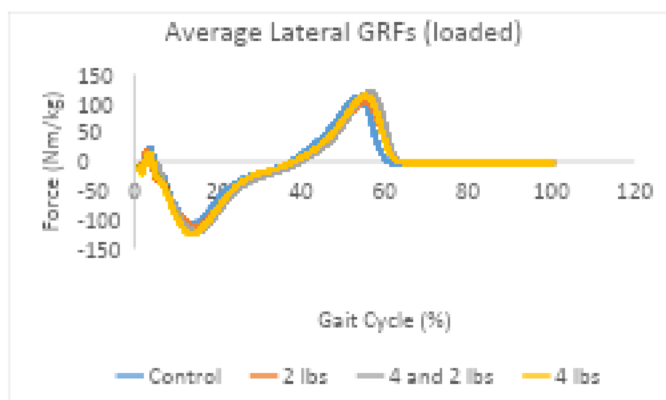


Figure 3. Ground reaction forces for the limb with the greatest load (loaded) and the limb with the lesser, or no, load (unloaded).



Averages were calculated based on 101 frames of normalized data.

4and2lbs trials ($P = 4.53*10^{-48}$). No significant differences were found for hip internal and external rotation or ground reaction forces (Fig. 3). However, from a qualitative standpoint, after graphing hip, knee, and ankle angles in MATLAB, there only appears to exist visible variation for the hip angles (Fig. 4).

IV. Discussion

The goal of this pilot study was to investigate whether there would exist significant differences in gait biomechanics and GRFs when placing small, lateral loads at the hips. Contrary to the hypothesis, significant differences were only observed for contralateral pelvic drop and trunk excursion in the frontal plane, not hip internal and external rotation, nor the peak GRFs between the weighted and unweighted limbs. Furthermore, it was thought that there would be greater observed compensatory adjustments that result from a larger weight difference between the two limbs.

Figure 2. Contralateral pelvic drop and trunk excursion. Averages were based on 101 frames of normalized data.

Table 1. Kinematic and Kinetic Variables of Interest.

Variable	Control, Mean ± SD	2lb Load, Mean ± SD	4lb Load(r) + 2lb Load(l), Mean ± SD	4lb Load, Mean ± SD	p-value
Pelvic drop excursion	4.0 ± 0.51	3.7 ± 0.1	4.5 ± 0.4	4.7 ± 0.7	$0.09*10^{-55}$
Loaded hip maximum internal rotation	10.9 ± 1.0	10.9 ± 1.5	10.1 ± 1.0	10.4 ± 0.3	0.71
Unloaded hip maximum internal rotation	12.5 ± 1.5	9.4 ± 1.1	10.2 ± 1.6	9.0 ± 0.9	0.77
Loaded hip maximum external rotation	28.1 ± 1.0	27.1 ± 0.4	29.1 ± 0.9	27.1 ± 0.7	0.71
Unloaded maximum hip external rotation	30.5 ± 0.4	27.8 ± 1.4	29.5 ± 0.8	28.5 ± 0.5	0.77
Trunk excursion (frontal plane)	11.1 ± 0.0	8.3 ± 2.5	6.6 ± 1.4	4.2 ± 0.7	$4.53*10^{-48}$
Peak GRF loaded limb	110.4 ± 0.7	10.5.9 ± 0.3	121.2 ± 4.5	113.5 ± 4.2	0.99
Peak GRF unloaded limb	115.2 ± 8.4	123.8 ± 0.9	131.2 ± 8.9	121.1 ± 3.0	1.0

**means and standard deviations were calculated based on 101 frames of data normalized to body weight

However, for pelvic drop excursion both the Control-2lbs and the Contrl-4lbs trials showed significant differences. Additionally, for trunk excursion, differences were observed in Control-4lbs and the Control-4and2lbs trials. For the remainder of the kinematic and kinetic variables, no differences were observed, suggesting that the joints that do experience compensatory movements are likely closer in location to the placement of the weight, at least under relatively low-weight, asymmetrical conditions. Further, the experimentally induced loads were found to be insufficient to produce changes in hip internal rotation, hip external rotation, and peak GRFs during walking in both the limb that had a heavier load added and both the limb that had little or no additional weight. Therefore, it is possible that the decrements in trunk excursion associated with greater amounts of asymmetry were not large enough to result in altered hip rotations during walking. However, a reduced trunk excursion may have biomechanical implications for more dynamic movement tasks such as running and jumping that place the muscles under greater demand. Normally the trunk moves side to side in the gait cycle

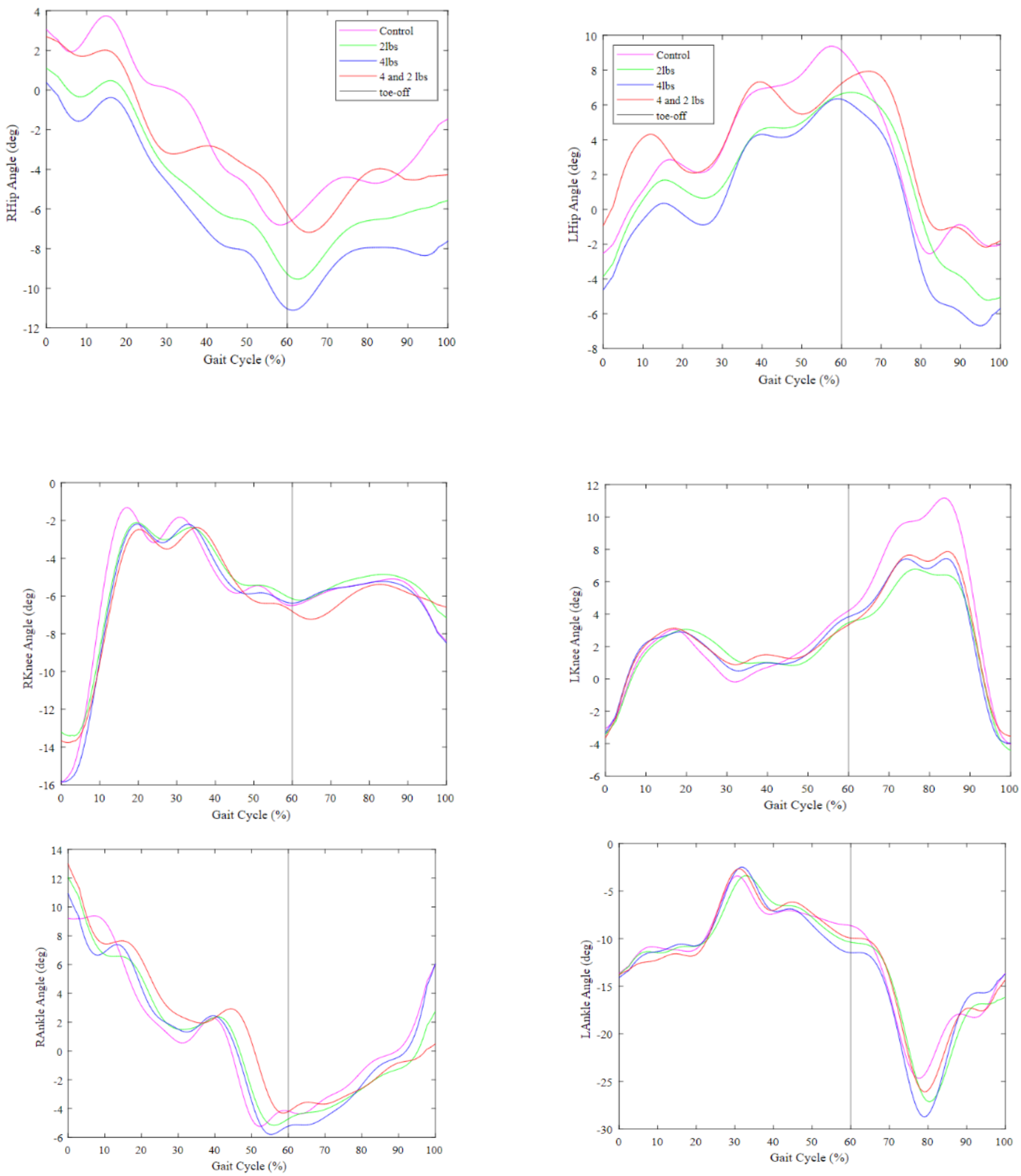


Figure 4. Loaded (right) and unloaded (left) hip, knee, and ankle angles.

(coronal plane) and aligns over each leg during its stance phase. This might be because of the need for support of the trunk during unilateral stance. It has been reported that the trunk moves towards the weight-bearing leg in normal gait at initial contact and then away from that side at terminal stance [16]. However, in our study, there was significant coronal asymmetry that resulted in excursion becoming more and more reduced during the full gait cycle, as the load was increased on the right hip. This may be attributed to an altered strategy of the trunk to lengthen to support balance as the person commences and completes swing on the side holding greater loads, or a collapse of trunk stability during stance on the side of the body withstanding more load. Though the participant also utilized little to no arm swing throughout the gait cycle, this is likely not the causal explanation for reduced trunk excursion. Previous studies have shown that there seems to exist no change in trunk angular momentum or trunk stability with different arm-swinging conditions [17, 18]. Thus, it is more likely that reduced trunk excursion is linked more so to

altered balance strategies. Yet, this data may also only hold true for female subjects. Nowicki (2005) discussed the importance of considering male and female body types in duty belt selection and equipment placement, specifically noting that female officers may need to keep their belts at slightly different positions than larger male officers with beltlines that can accommodate more items [19]. However, within the limitations of this pilot study, it is concluded that even small lateral load asymmetries placed at the hips result in significant biomechanical differences in gait and ground reaction forces, specifically at joints near the location of the load. A more extensive study involving a larger sample size is needed to fully validate the findings of this study. Further research is also required to determine whether a similar relationship between male subjects and frontal-plane gait biomechanics exists in adults when asymmetrically loaded.

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