ERGONOMIC ANALYSIS OF ARMY BACKPACK DESIGNS: BACK AND SHOULDER STRESSES AND THEIR IMPLICATIONS

by

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STATEMENT OF DISSERTATION APPROVAL

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ABSTRACT

Over the years, the soldier's load has increased; weapon system improvements and the need for increased protection and firepower require individuals to carry more equipment. Although the current army field manual provides recommended guidelines for a soldier's load per operation, soldiers typically carry loads exceeding the recommended guidelines. The overall effect of these heavy loads on the soldier's body and the impact on the soldier's performance is still uncertain.

In this study, we analyzed the existing and proposed Korean army backpack designs and determined how each of the designs impacts stress on the soldier's upper body.

Twenty healthy male subjects participated in this study. Subjects were selected from among University of Utah students who have not experienced or have fully recovered from discomfort, injuries, or disorders that could affect normal gait.

Each trial had 3 repetitions. The independent variables being controlled were surface types and orientations, backpack types and loads, and marching speed.

While each subject walked on the tracks with or without a backpack, three-dimensional motion data and analog data (EMG, load cell) were collected with 16 Optitrack V100:R2 cameras, AMASS software, and LabVIEW. The captured data were then processed with Visual3D, Vicon Nexus, and MATLAB software.

Using inverse dynamics and recorded erector spinae electromyography (EMG) data, force on the L5/S1 disc was estimated using the proposed biomechanical model. Shoulder force data was measured from customized load cells integrated into the shoulder straps of the backpacks.

Upper body segments exhibited greater deviations from neutral positions (i.e., greater thorax flexion, greater thorax lateral flexion, and more pelvic anterior tilt) when carrying a backpack than under normal walking conditions. These deviations resulted in increased shoulder tension, which, in turn, increased compressive and shear forces on the L5/S1 disc.

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저의 논문을 위해 많은 시간을 할애하고 물심양면으로 도와주었던 유병주, 김성원, 김충신, 이상주 님께 진심으로 감사의 말씀을 드립니다. 잊지않겠습니다.

자랑스러운 나의 조국 대한민국과 육군에 이 논문을 바칩니다.

CHAPTER 1

INTRODUCTION

Problem Identification

The ROKA (Republic of Korean Army's) new backpack design project was initiated in August, 2010. The new design focuses on an increase in usability and a decrease in soldier fatigue resulting from carrying a heavy load. In this regard, the goal was to develop a modularized type of backpack as shown in Figure 1-1 (ROKA, 2010).

As with most soldiers operating on the ground, Korean soldiers are required to carry heavy loads during military operations and training. Over the years, the soldier's load has increased due to weapon system improvements, and the need for increased protection and firepower, which require individuals to carry more equipment (Knapik et al., 1996). The soldier's backpack load is generally considered one of the most significant factors in military operations, and therefore in the weapon system R&D (research and development) field.

Combat loads also play a significant role in determining the continuous operations capability of soldiers and troops. This is because even a minor injury created by the loads might cause noncombat losses during continuous military operations. Knapik examined injuries associated with maximum effort marching training. During the observed training, 333 soldiers carried 46 kg loads over a 20 km course. Of these soldiers, 24 % had injuries such as a foot blisters, back pain, ankle sprains, and so on. Among these injuries, foot blisters (35 %) and back problems (23 %) were reported as the most common (Knapik et al., 1992). These injuries may seem to be minor problems, but, in this study alone, they resulted in 44 days of limited duty. This represents a huge noncombat loss to the commander.

Many factors influence a soldier's load carrying ability. These factors include load weight, marching speed, type of terrain, load distribution, medical condition, and so on (Kinoshita, 1985; Pandolf et al., 1977; Patton et al., 1991). Although current army field manual recommends guidelines for the soldier's load on each operation (USFM21-18), soldiers typically carry loads exceeding the recommended guidelines (Knapik et al., 1997), and the overall effect of these heavy load on the soldier's body and the impact on the soldier's performance is still uncertain.

General Statement of Research Required to Address the Problem

In military operations, backpacks are a basic load carriage method for infantry soldiers. Some studies related to load carriage have been performed. It is difficult, however to find research related to loads, load configurations, and operational surfaces specific to military personnel in general, much less related to the Korean military.

In this study, we will analyze the existing and proposed Korean army backpack design and determine how each of the designs impacts the stress on the soldier's upper body. Back compressive forces and shoulder reaction force will be quantified for the gait cycle. This empirical research also has the potential to expand the scope of the Korean army's weapon system design process and methodology.

Figure 1-1. Modularized Design Concept of the New ROKA Backpack

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CHAPTER 2

METHODS

Subjects

Participants were carefully selected, with similar characteristics, from a healthy young population who fit the current military soldiers' recruiting standard. Twenty healthy male subjects participated in this study. They were selected from University of Utah students who have not experienced or fully recovered from any discomfort, injuries, or disorders that may affect normal gait. The anthropometric selection criteria were limited to ages of 18 to 30, height of 161 to 195 cm, and weight of 55 to 87 kg to replicate the current military soldiers' recruiting standard. Table 2-1 shows recruited participants' anthropometric data, based on our participant selection criteria. All participants had enough time to review the IRB consent form approved by University of Utah Institutional Review Board. They also were notified that they could drop study participation at any time during the trials if they felt uncomfortable.

Experimental Design

The independent variables controlled for this study were: surface types and orientation, backpack types and loads, and marching speed. Specifically they were: 2 surface types (hard / sand), 2 surface orientations (flat / slope), 3 loading types (MOLLE /

ALICE / no backpack), and 2 speeds (self-paced / 4 km/h). Each trial had 3 repetitions. Thus, the total number of trials per subject was 72.

surface composition (2) x slant (2) x backpack (3) x speed (2) x 3 times = 72

A randomized block design was used, where the track (surface composition and side slope) was the blocking parameter, meaning all necessary trials were performed for that specific blocked condition in succession.

Data Collection Protocol

While each subject walked on the tracks with or without a backpack, three-dimensional motion data and analog data (EMG, load cell) were collected with 16 Optitrack V100:R2 cameras and AMASS software, and LabVIEW. The captured data were processed with Visual3D, Nexus and MATLAB software.

Data collection per each participant took approximately half a day (4 to 5 hours). During this time the subject was asked to walk down a 24 ft. walkway repeatedly until 72 successful trials were collected. Duration per each trial was 5 seconds. We asked and visually checked each participant's physical condition between each trial to minimize the effect of fatigue during data collection. Actual total walking distance and time was approximately 0.67 miles and 40 min., respectively. After collecting 3 trials per condition, each participant was provided with sufficient recovery time. When the track was sloped to the side, measurements were taken when the right foot was always in downslope and left foot was always in upslope.

For static trials, reflective markers were attached bilaterally to the subjects at the following locations:

1. Pelvis: Right ASIS (anterior superior iliac spine), left ASIS, right PSIS (posterior superior iliac spine), and left PSIS
2. Thorax: RSHO (right shoulder), LSHO (left shoulder), C7 (7th cervical vertebra), STRN (sternum), XIPH (xiphoid process)
3. Backpack: right center, left center
4. Lower limb: lateral femoral condyle, medial femoral condyle, lateral malleolus, medial malleolus, calcaneus, between the second and third proximal metatarsal heads, head of 5th Metatarsus

For dynamic walking trials, because backpacks block PSIS markers all the time during dynamic trials, virtual PSIS markers were introduced and measured using thigh clusters (Figure 2-1).

A static trial was captured for 6 seconds for each subject in order to calibrate the marker set and to create a model.

The static marker set, as mentioned above, was used to find the hip joint center locations using the relationship between two ASIS and two PSIS markers and thigh clusters before starting the data collection.

Then, we identified the virtual hip joint center from the relationship between the PSIS marker, the virtual hip joint center location from thigh clusters and two ASIS markers. By this process, the virtual PSIS markers could always be tracked by the other markers even if that marker is missing or blocked. These markers describe the location of each body segment at any point in time for calculating joint positions, velocities, and accelerations.

After the participant is equipped with the markers and calibration measurements have been taken with the computer based motion analysis program, the researcher asked the participant to walk down the track (Figure 2-2).

Two force plates (OR6-5-1000 & OR6-7, AMTI, Watertown, MA) measured ground reaction forces on each foot while walking on the track (Figure 2-3).

Using inverse dynamics and recorded erector spinae electromyography (EMG) data, force on L5/S1 disc was estimated using the proposed biomechanical model. Shoulder force data was measured from customized load cells integrated into the shoulder straps of the backpacks (Figure 2-4).

The load of each backpack was fixed at 28 kg based on Korean army backpack design guideline (ROKA, 2010). The empty MOLLE pack weighed 1 kg and the ALICE was 400 g. Thus, total weights of the backpacks were 29.0 kg (MOLLE) and 28.4 kg (ALICE), respectively.

Participants were asked to walk at two different walking speeds, which are self-paced and controlled speed at 4 km/h. Self-paced speed was freely chosen by the participant as their normal walking speed. For controlled speed, they followed the guiding flag that was moving constantly at 4 km/h at their eye height.

Their self-paced walking speed and controlled speed resulted in 3.99 km/h and 4.39 km/h, respectively, and the speeds were significantly different ($p<.005$).

Controlled speed, at 4 km/h, resulted in increased cadence, decreased double support time, and increased stride length. Walking at 4 km/h would thus be expected to increases fatigue. Its effects were identified and discussed in further detail in the lower limb analysis, which is the other part of our research project.

All participants wore military issued boots to control the effects of footwear in our study (Figure 2-5).

Statistical Analysis

Analysis of variance (ANOVA) was mainly used to analyze the data. Tukey LSD was used for post-hoc analyses when necessary. The level of significance was set as 0.05 for all statistic analysis. SPSS 18.0 (IBM Corporation, Armonk, NY) was used for analysis.

Table 2-1. Average Participants' Anthropometric Data

	Age (yrs.)	Height (cm)	Mass (kg)
Mean	25.1	175.6	74.9
SD	3.6	4.6	7.7

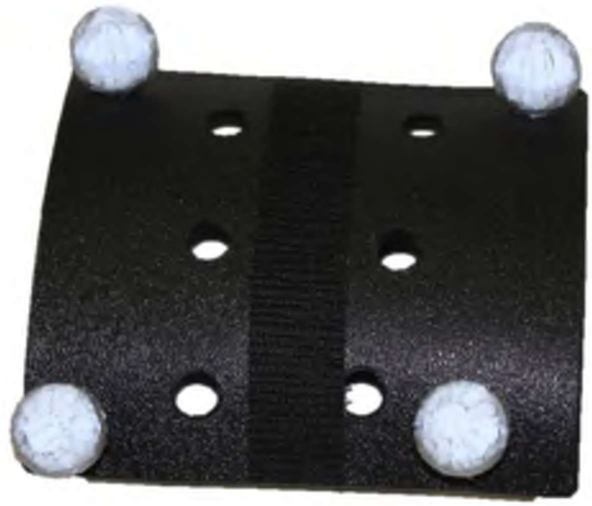

Figure 2-1. Marker Cluster

Figure 2-2. Dynamic Trial with Reflective Marker Set

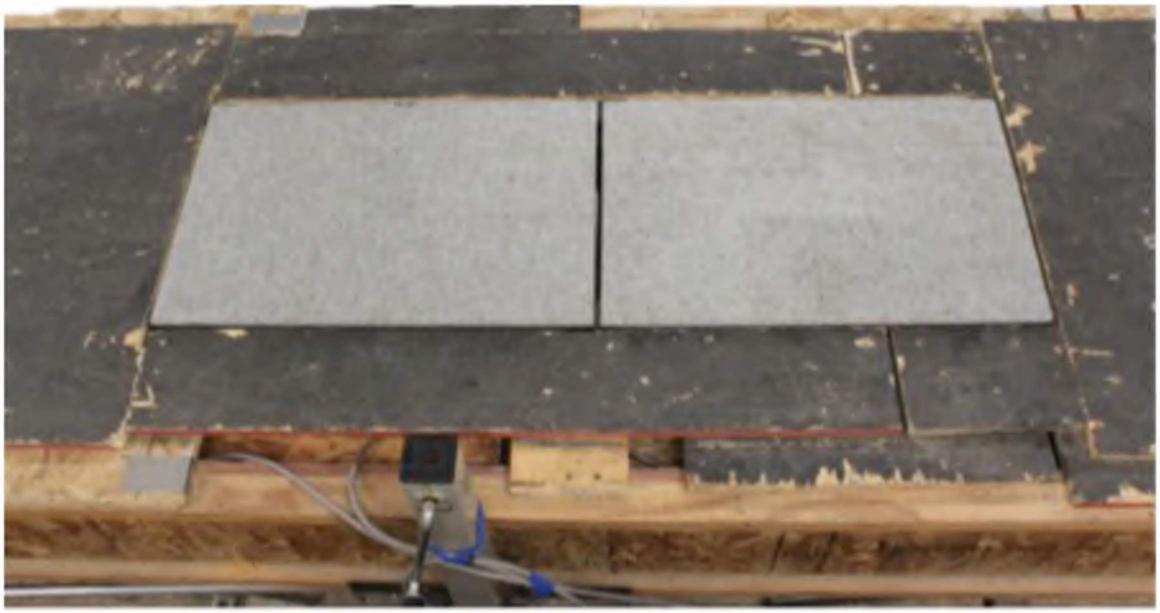

Figure 2-3. Force Plates Setup

Figure 2-4. Backpack (ALICE) and Load Cell Setup

Figure 2-5. Military Issued Boots

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CHAPTER 3

THE EFFECTS ON UPPER BODY KINEMATICS WHEN WALKING WITH A MILITARY BACKAPCK

The purpose of this chapter is to identify the effects of military backpack carriage on soldiers' upper body kinematics, focusing on thorax and pelvis motions. Statistical significances were identified during loaded walking compared to normal walking conditions, such as increased thoracic flexion and pelvic tilt in the sagittal plane. We also identified the potential risk of backpack carriage from deviated postural motion due to backpack load.

Introduction

Current significant operational issues in the Korean military relate to MOUT (military operation on urban terrain) and desert operations. In modern warfare, most urban operation fields are composed of hard surfaces. The soldier's performance on sandy surfaces has become more important due to the frequency of desert operations. It is also important to have some understanding of the effect of load carrying on laterally uneven surfaces.

Many studies related to load carriage have been done on even surfaces (Birrell et al., 2009; Quesada et al., 2000; Vacheron et al., 1999). Majumdar evaluated lower limb kinematic responses according to the gait cycles, and noted the necessity of modifying the existing Indian backpack design, especially for use in low intensity conflict environments. In his research, 10 male infantry soldiers were asked to walk a track with the backpack on an even surface. He asked subjects to walk at a self-selected pace and observed that the forced marching speed causes abnormal gait patterns among soldiers (Majumdar et al., 2010).

It is often required, however, for soldiers to walk at a specific speed, even though this activity might expose them to higher injury risks. This is because a specific marching speed may be essential to the commander's tactical operational needs. For instance, if a commander has information about the distance to opposing troops, a specific marching speed may be necessary to engage the enemy at a selected point. Currently the US army field manual identifies a typical marching speed of 4 km/h (USFM21-18), which is the same marching speed used by the Korean army.

Merryweather analyzed the effects of walking on ballast (small and large gravel) and left/right slanted surfaces. In his study, a 3D (3-dimensional) motion analysis system was used to capture and analyze each subject's specific motion characteristics while each subject walked on the tracks filled with ballast and on a hard surface. Force plates measured the force, moment and the foot/surface interface, and using inverse dynamics the forces and moments at the ankle and knee were calculated. He found that both surface types (hard vs. ballast) and slant (even vs. left/right) had an effect on the knee joint forces (Merryweather, 2008).

Method

Equipment

Two walkway tracks with two force plates were designed for a previous research project (Merryweather, 2008) in which lower limb biomechanics were analyzed for subjects walking on slanted and level railroad ballast. It was modified to fit the current study. One track was built for the sand surface and the other track was built for the hard. Both tracks have a height adjustable feature to simulate side slope conditions up to 15 degrees. The sand was selected based on the guidance of a geological expert and a former resident of Iraq to best simulate the desert environment in the Middle Eastern region (Figure 3-1). AMASS® motion capture software with 16 cameras was set up around the tracks to capture the motion data. Figure 3-2 shows the camera setup schematically.

Marker Setup

For thorax modelling, RSHO (right shoulder), LSHO (left shoulder), STRN (sternum), XIPH (xiphoid process), and C7 (7^{th} cervical vertebra) markers were used. RAS (right anterior superior iliac spine), LAS (left anterior superior iliac spine), RPS (right posterior superior iliac spine), and LPS (left posterior superior iliac spine) markers were attached for pelvis modelling. Figure 3-3 shows the anatomical location of each marker. For dynamic trials, RSHO, LSHO, RPS, and LPS markers were removed after static capture.

Data Analysis

The kinematic data were collected with a sampling frequency of 100 Hz, and then filtered with a Butterworth low pass filter at 6 Hz cut-off frequency in Visual3D when processing.

The collected data were analysed using SPSS (Ver.18.0, IBM Corporation, Armonk, NY) with the significance level of 0.05. Thoracic flexion (sagittal), thoracic lateral extension (coronal), thoracic rotation (transverse), pelvic tilt (sagittal), pelvic obliquity (coronal), pelvic rotation (transverse), and their relative motion in each plane were analyzed using MANOVA and post-hoc test.

To maintain consistency, gait cycle was normalized from 0 % (left heel strike) to 100 % (left heel strike).

Results

Thoracic Flexion with regard to Global Coordinate System in Sagittal Plane

Figure 3-4 shows mean thoracic flexion angles; positive angles mean forward flexion in the sagittal plane.

From MANOVA, thoracic flexion angles showed statistical differences. Higher thoracic flexion was found on sand than on the hard surface (p<.001; Figure 3-5); on the sloped surface than on the flat surface (p<.001; Figure 3-6); at self-paced speed than at 4 km/h (p=.01; Figure 3-7); and with loaded walking than with the no loading condition (p<.001; Figure 3-8). In addition, interaction between surface type (hard, sand) and speed (self-paced, 4 km/h) was significant (p=.019; Figure 3-9).

Thoracic Lateral Bending with regard to Global Coordinate System in Coronal Plane

Figure 3-10 shows mean thoracic flexion angles; positive angles mean forward flexion in the sagittal plane.

Thoracic lateral motion showed statistical differences. Positive angles in the figures mean thoracic flexion (left shoulder up), and negative angles represent thoracic extension (right shoulder up) in the coronal plane. Higher thoracic extension was found when loaded than with no load (p<.001; Figure 3-11); and at self-paced speed than at 4 km/h (p=.005; Figure 3-12). No interaction was significant.

Thoracic Rotation with regard to Global Coordinate System in Transverse Plane

Figure 3-13 shows mean thoracic rotation angles in the transverse plane.

From MANOVA, thoracic transversal rotation showed statistical differences. Positive angles in the figures mean thoracic anterior rotation (right shoulder anterior), and negative numbers represent thoracic posterior rotation (right shoulder posterior) in the transverse plane. Greater thoracic anterior rotation was found on the sloped surface than on the flat surface (p=.001; Figure 3-14); at self-paced speed than at 4 km/h (p=.001; Figure 3-15); and with loaded walking than with the no loading condition (p=.001; Figure 3-16). Interaction between slope and backpack was significant (p=.039; Figure 3-17).

Pelvic Tilt with regard to Global Coordinate System in Sagittal Plane

Figure 3-18 shows mean pelvic tilt angles in the sagittal plane.

Pelvic sagittal motion showed statistical differences. Positive angles in the figures mean pelvic anterior tilt in the sagittal plane. Higher pelvic anterior tilt was found on the sand surface than on the hard surface ($p<.001$; Figure 3-19); and with loaded walking than with the no loading condition ($p<.001$; Figure 3-20). Interaction between speed and slope was significant ($p=.025$; Figure 3-21).

Pelvic Obliquity with regard to Global Coordinate System in Coronal Plane

Figure 3-22 shows mean pelvic obliquity in the coronal plane.

Pelvic obliquity showed statistical differences. Positive angles in the figures mean right side lifting in the coronal plane and vice versa. Higher pelvic obliquity amplitude was found on the sand surface than on the hard surface ($p<.001$; Figure 3-23); and on the flat surface than on the sloped surface ($p<.001$; Figure 3-24). Interaction between surface and backpack was significant ($p=.016$; Figure 3-25).

Pelvic Rotation with regard to Global Coordinate System in Transverse Plane

Figure 3-26 shows mean pelvic rotation in the transverse plane.

Pelvic transversal motion showed statistical differences. Higher pelvic rotation was found on the sloped surface than on the flat surface ($p<.001$; Figure 3-27); and with normal walking than while loaded ($p=.004$; Figure 3-28). Interaction between surface and speed was significant (p=.012; Figure 3-29).

Thorax-Pelvis Relative Motion in Sagittal Plane

Figure 3-30 shows mean thorax-pelvis motion in the sagittal plane.

Relative thoracic flexion angle due to pelvic tilt showed statistical differences. Higher thorax-pelvis flexion was found on the sloped surface than on the flat (p<.001; Figure 3-31); with loaded walking than with no load (p<.001; Figure 3-32); and at self-paced speed than at 4 km/h (p=.001; Figure 3-33). Significant interaction was identified between surface and backpack (p=.009; Figure 3-34) and between slope and surface (p<.001; Figure 3-35).

Thorax-Pelvis Relative Motion in Coronal Plane

Figure 3-36 shows mean thorax-pelvis motion in the sagittal plane.

Relative thoracic lateral motion due to pelvic obliquity showed statistical differences. Higher lateral motion (right shoulder up) was found with self-paced speed than at 4 km/h (p=.034; Figure 3-37); and with loaded walking than with no load (p=.001; Figure 3-38). There was no significant interaction.

Thorax-Pelvis Relative Motion in Transverse Plane

Figure 3-39 shows mean thorax-pelvis relative rotation in the transverse plane.

Relative thoracic transversal motion due to pelvic rotation showed statistical differences. Higher lateral motions (right shoulder anterior) were found with self-paced speed than at 4 km/h (p<.001; Figure 3-40); and with loaded walking than with normal walking ($p<.001$; Figure 3-41). There was significant interaction between slope and backpack ($p=.041$; Figure 3-42).

Discussion

Effects of Load Carriage on Upper Body Movement Profiles

Pairwise comparison (Tukey LSD) was performed to investigate the effects of each loading condition (no load, MOLLE, ALICE). Table 3-1 summarizes the post-hoc analysis results.

From kinematic analysis of the upper body, we found there were significant differences between normal walking (no load) and loaded walking. Upper body segments were exposed to more deviations, such as greater thorax flexion, greater thorax lateral flexion, and more pelvic anterior tilt, when carrying a backpack from normal walking condition.

Figure 3-43, Figure 3-44, and Figure 3-45 show sagittal motion profiles of the thorax and pelvis.

Thoracic and pelvic movement in the sagittal plane is closely related with erector spinae muscle contraction (Crosbie et al., 1997). Crosbie et al. addressed that "the spinal movements associated with walking are linked to the primary motions of the pelvis and the lower limbs"(1997). From our study, we found that peak thoracic flexion occurred at 44 %, and then peak pelvic anterior tilt followed at 54 % of gait cycle.

By combining the thoracic-pelvic motion with erector spinae muscle contraction timing, we could possibly build upper body stabilization mechanism when carrying a backpack. The relationship will be further discussed in Chapter 4.

In addition, increased flexion angle would change L5/S1 disc angle, and the angle is one of main contributors for estimating forces on the lumbosacral disc. The effects will be discussed further in Chapter 6.

Figure 3-46, Figure 3-47, and Figure 3-48 show coronal motion profiles of the thorax and pelvis.

We found asymmetrical profiles of thorax movement in the coronal plane which agreed with previous studies (Bartonek et al., 2002; Nguyen et al., 2004). The asymmetric profile of thoracic motion was more evident when carrying backpacks. It might increase the possibility of back pain or injuries from unbalanced movement of the spine. However, the potential risks remain unclear.

We identified that the right shoulder was lower than the left shoulder when walking without load. No other study explained the reason for the asymmetry, but one online source observes "I often notice that…dominant shoulder is lower than their recessive shoulder" (http://ericbeard.com). Given this observation, we infer that the dominant shoulder of most subjects is the right, however, further investigation is still required to understand the reason for the asymmetry.

With a load, the right shoulder was identified as being higher than the left shoulder. This might be explained from the higher tolerance of the right shoulder muscles given the assumption that the right side was the dominant shoulder in our study. Humans tend to optimize their behaviors for minimizing the possibility of injuries and fatigue. When they can control their walking speed (self-paced) with the backpack on, they might try to minimize shoulder muscle fatigue and protect the weaker (recessive) shoulder from injury or pain by lifting up their stronger (dominant) shoulder. At controlled speed, this tendency might be overwritten to catch up the speed. Thus, forced marching in the military might be riskier than casual walking or hiking. More research is needed in the future to prove the assumptions that we have made for this explanation.

Lateral movement of thorax is also related with shoulder reaction force profile. It will be discussed in Chapter 5.

Average lateral bending angles were within 2 degrees. When there was no load, the fluctuation in lateral motion was higher when walking at 4 km/h than at a self-paced speed. On the other hand, with a backpack load the magnitude was bigger when walking at self-paced speed than at 4 km/h.

Figure 3-49, Figure 3-50, and Figure 3-51 show profiles of transverse motion of the thorax and the pelvis. As we can see, the thorax rotated contralaterally to the foot on the ground.

There was a significant decrease in transverse pelvic rotation when carrying a backpack. LaFiandra et al. also found the decrease in their research (2003). They explained it from decreased stride length due to load carriage (LaFiandra et al., 2003). From lower limb study, which was another part of our project, significantly shorter stride length was identified with load. We could thus conclude that shorter stride length resulted in decreased amplitude of transverse pelvic rotation. Additional whole body analysis is essential in the future to investigate complex mechanisms and characteristics of human body motions.

Conclusion

In this chapter, we found that upper body segments were exposed to more deviations, such as greater thorax flexion, greater thorax lateral flexion, and more pelvic anterior tilt, when carrying a backpack. There were also higher thoracic flexion and pelvic anterior tilt in the sagittal plane when walking on the sand surface.

Asymmetrical profiles of thorax movement were identified in the coronal plane. The asymmetric profile of thoracic motion was more evident when carrying backpacks. This unbalanced movement of the spine might increase the possibility of back pain or injuries. However, the potential risks remained unclear.

Backpack carriage induced a significant decrease in transverse pelvic rotation, and this can be explained from decreased stride length due to the load carriage. Additional whole body analysis is essential, in the future, to investigate complex mechanisms and characteristics of human body motions.

Figure 3-1. Tracks and Force Plates

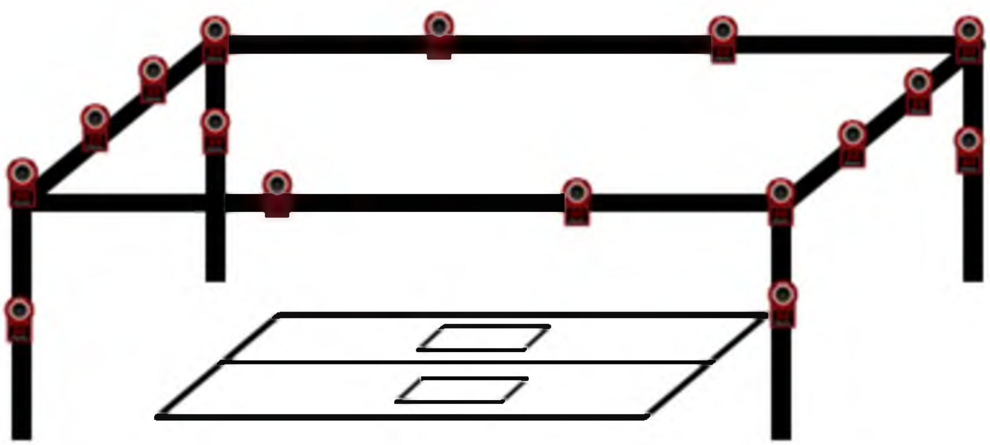

Figure 3-2. Schematic Drawing for Camera Setup

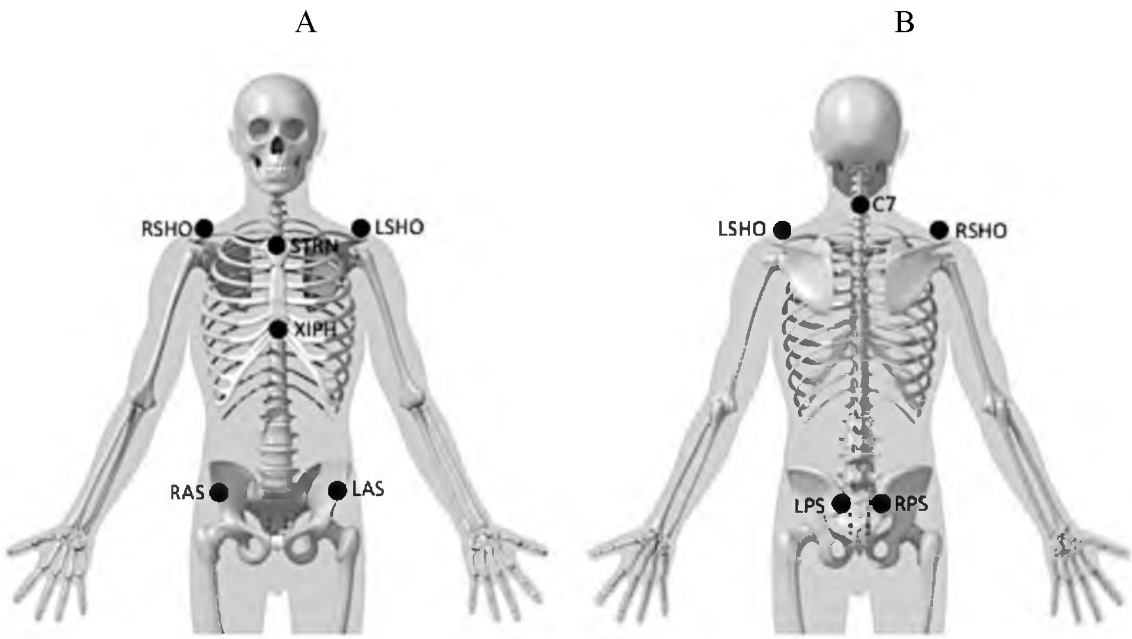

Figure 3-3. Marker Setup. A) Front; B) Back

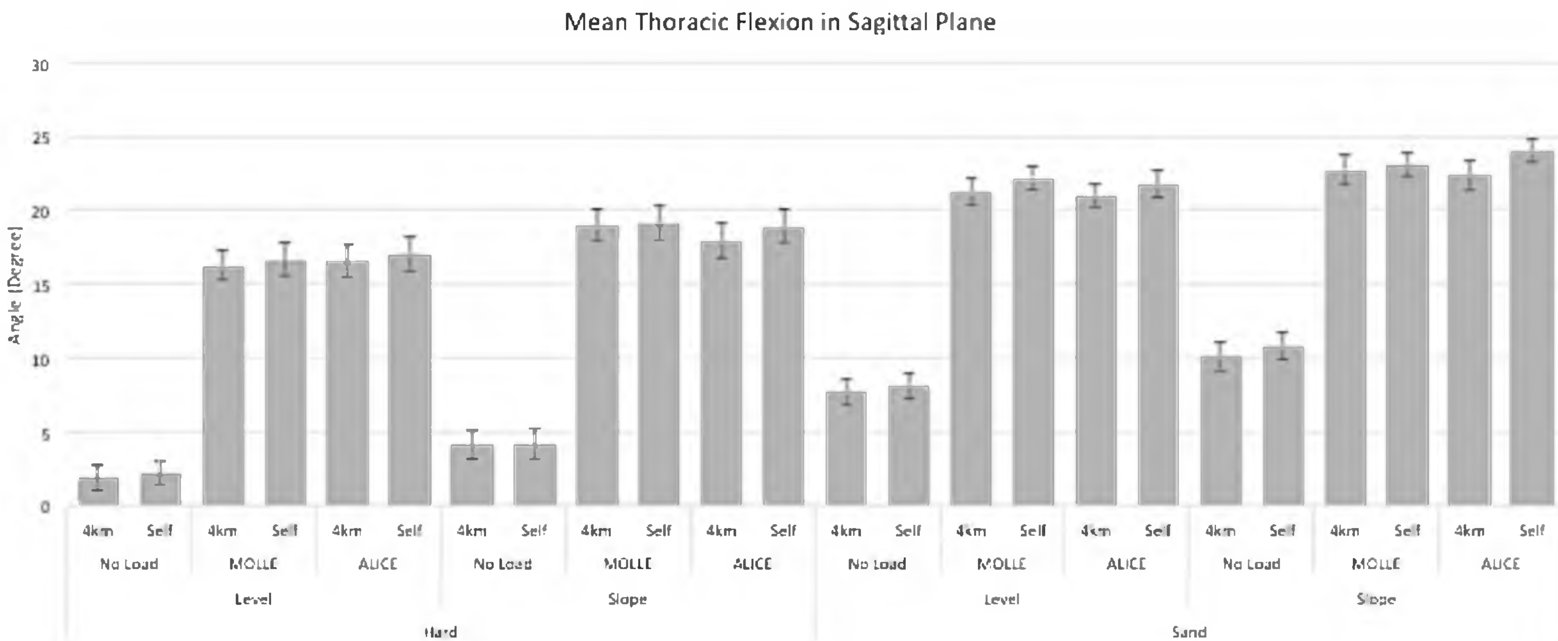

Figure 3-4. Mean Thoracic Flexion in Sagittal Plane

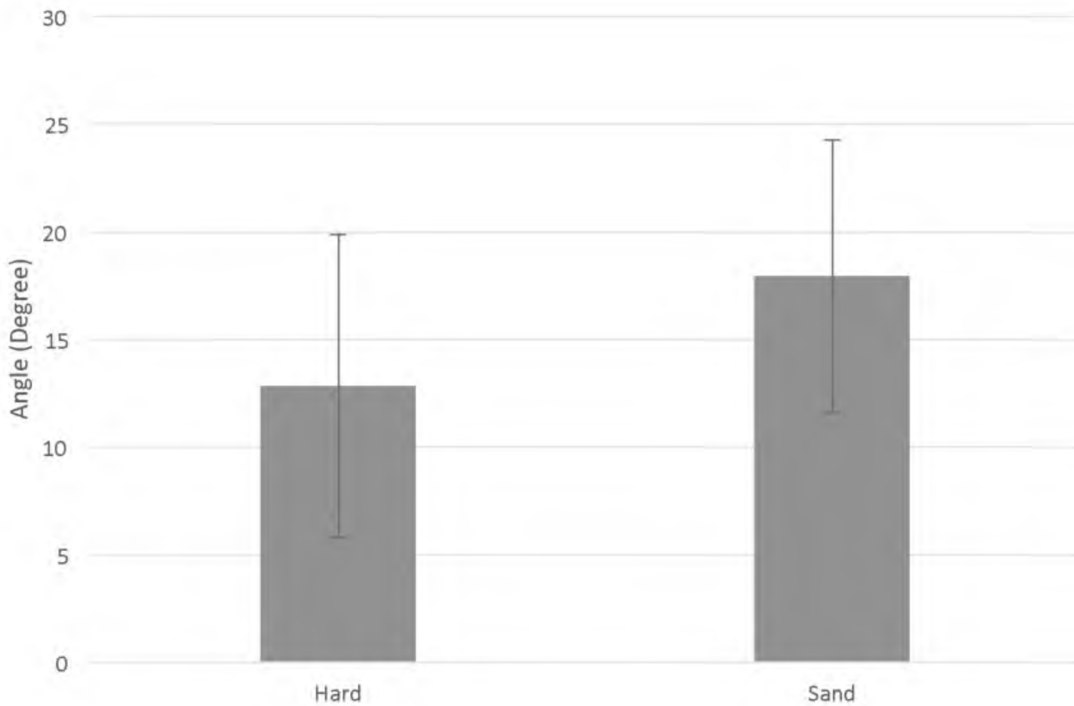

Figure 3-5. Thoracic Flexion by Surface ($p<.001$)

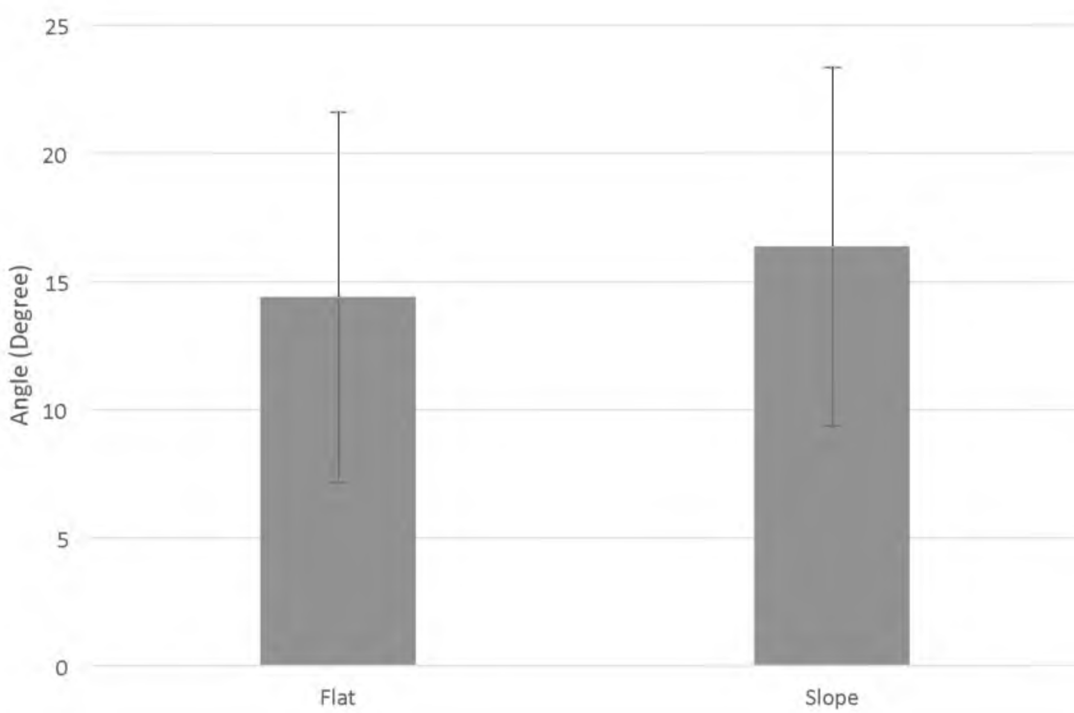

Figure 3-6. Thoracic Flexion by Slope ($p<.001$)

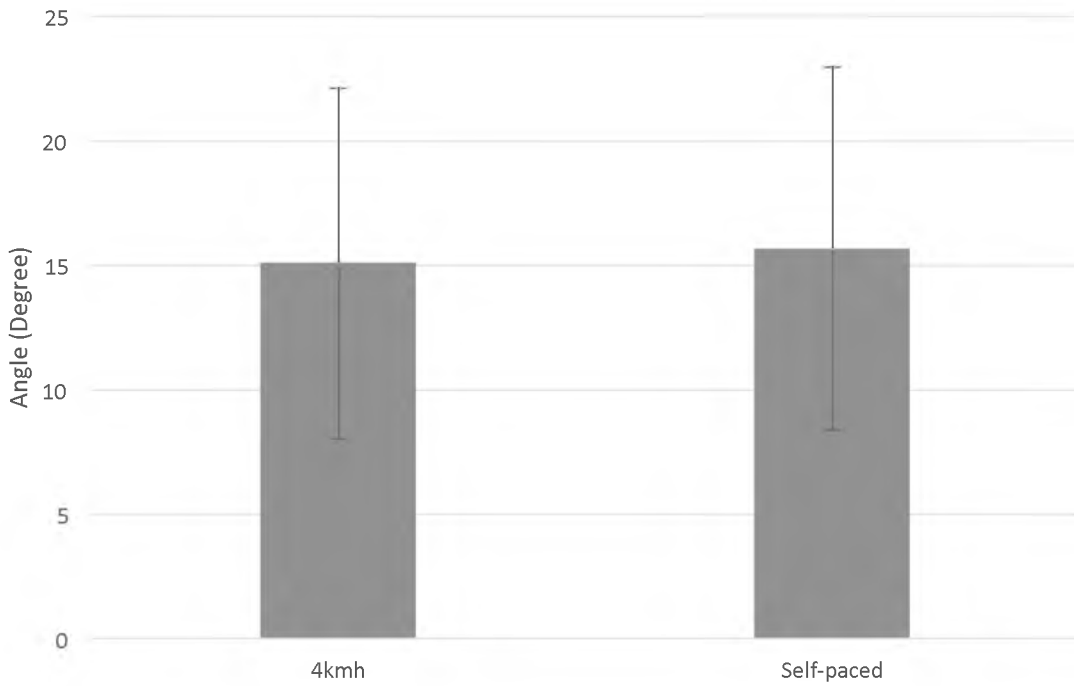

Figure 3-7. Thoracic Flexion by Speed (p=.01)

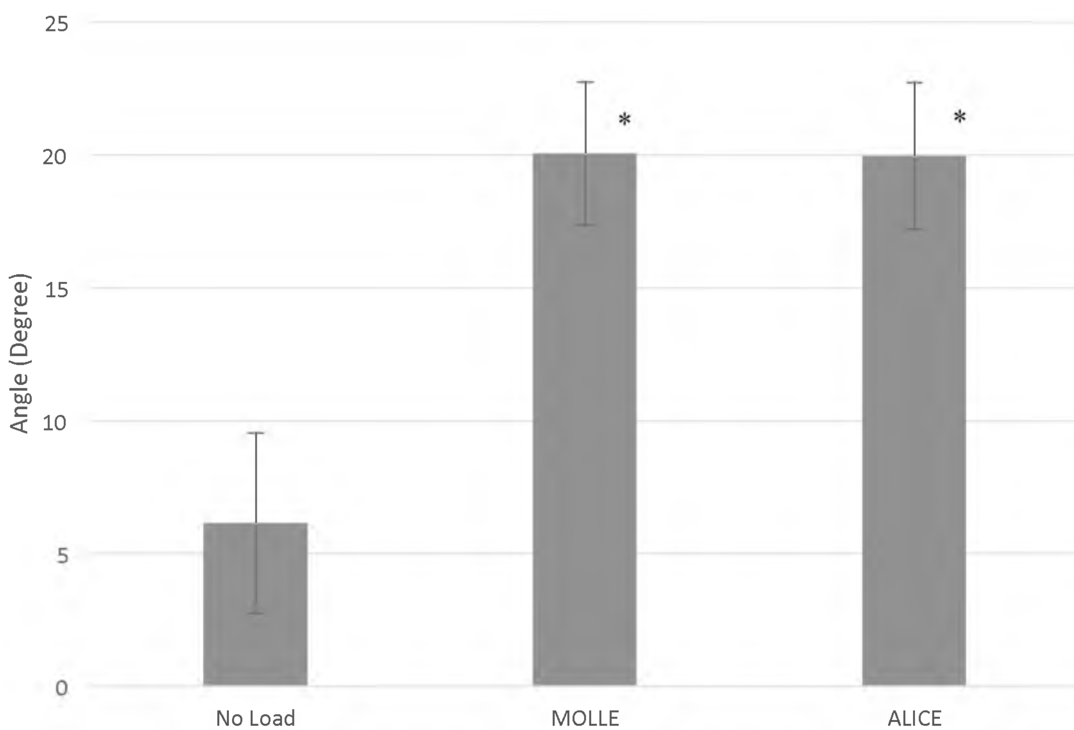

Figure 3-8. Thoracic Flexion by Backpack (p<.001)
* indicates statistical significance (p<.005) compared to No Load

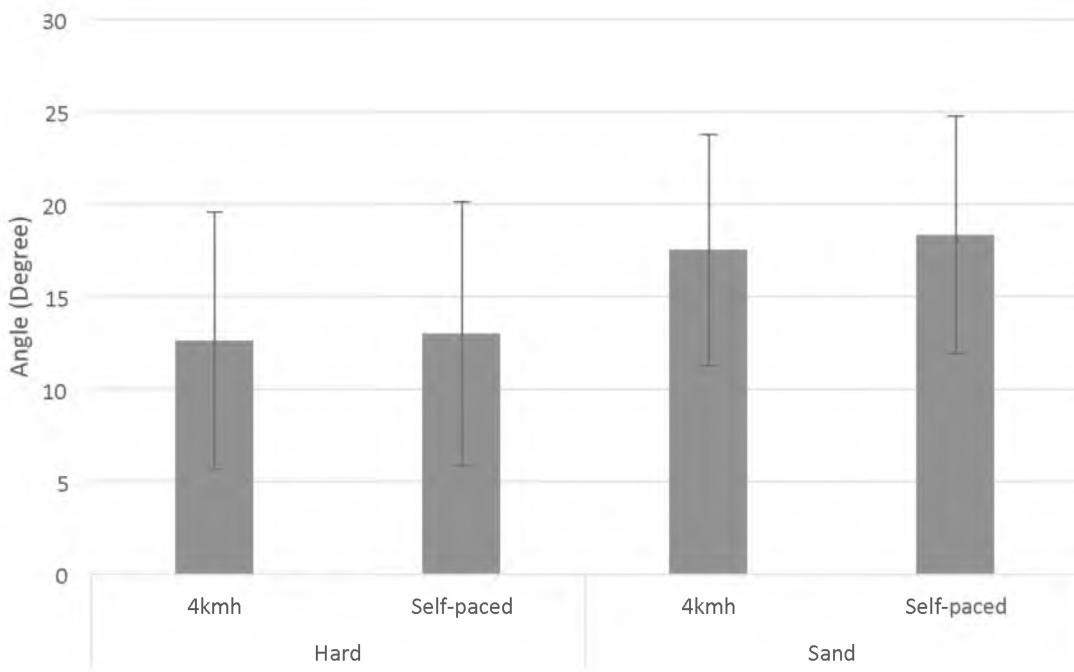

Figure 3-9. Thoracic Flexion by Surface and Speed (*p*=.019)

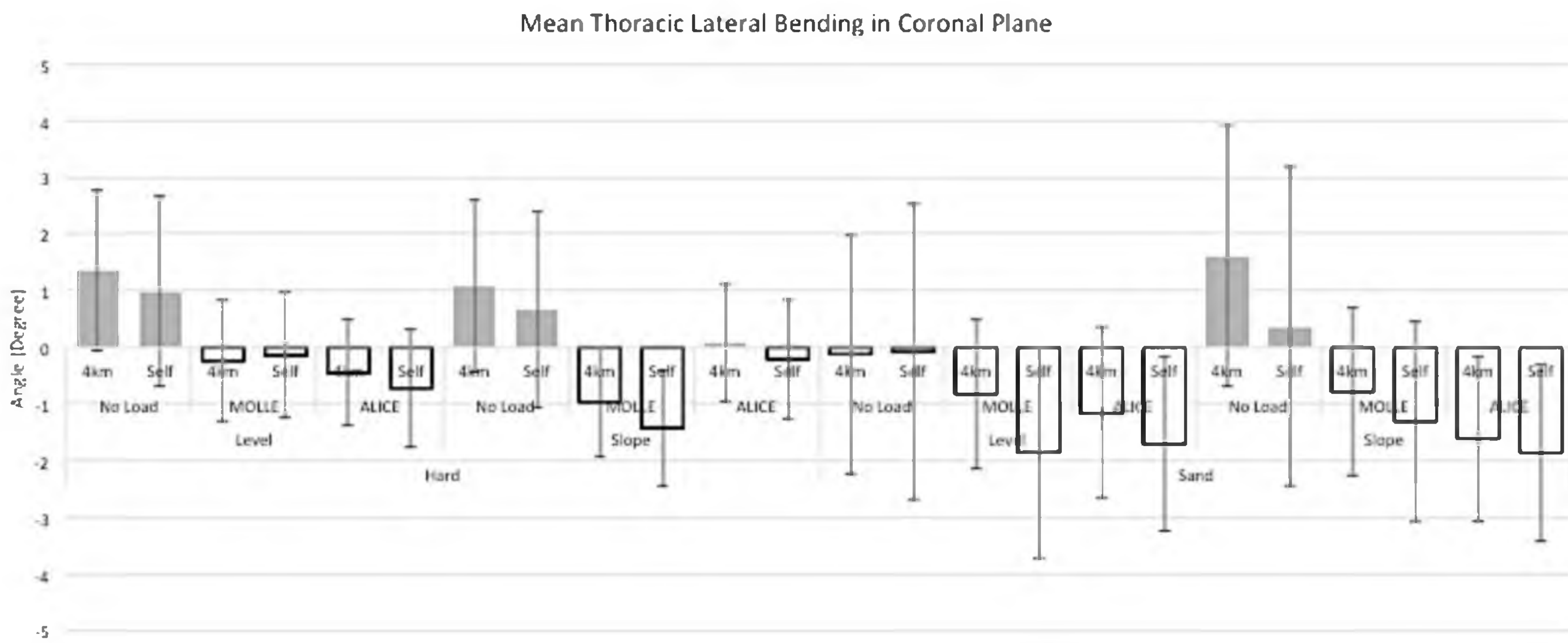

Figure 3-10. Mean Thoracic Lateral Bending in Coronal Plane

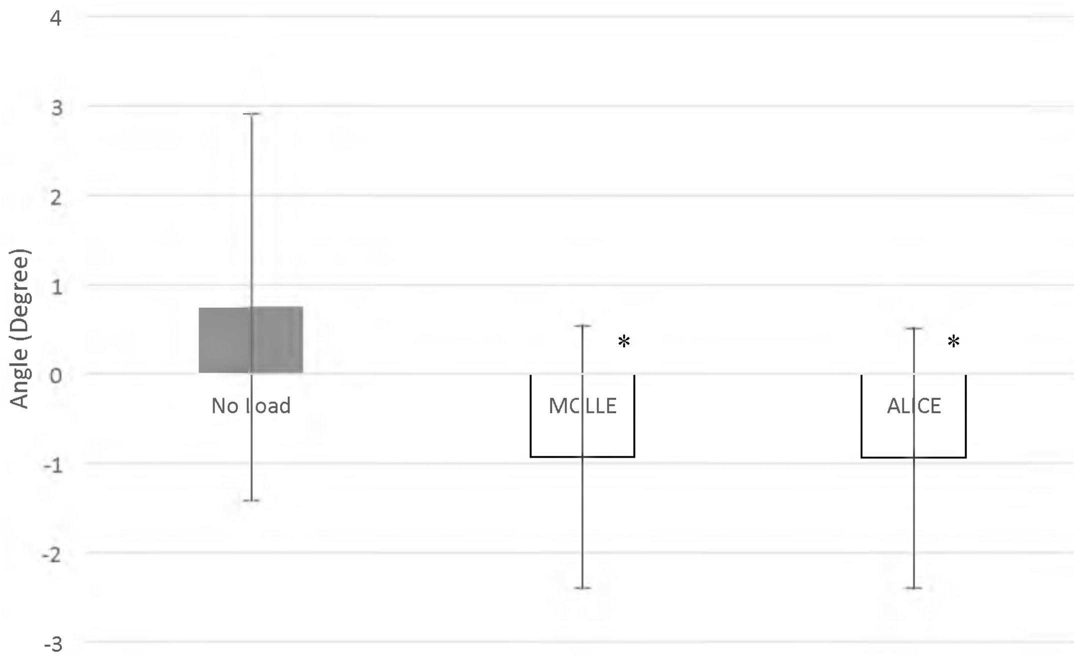

Figure 3-11. Thoracic Lateral Bending by Backpack ($p<.001$)
* indicates statistical significance ($p<.005$) compared to No Load

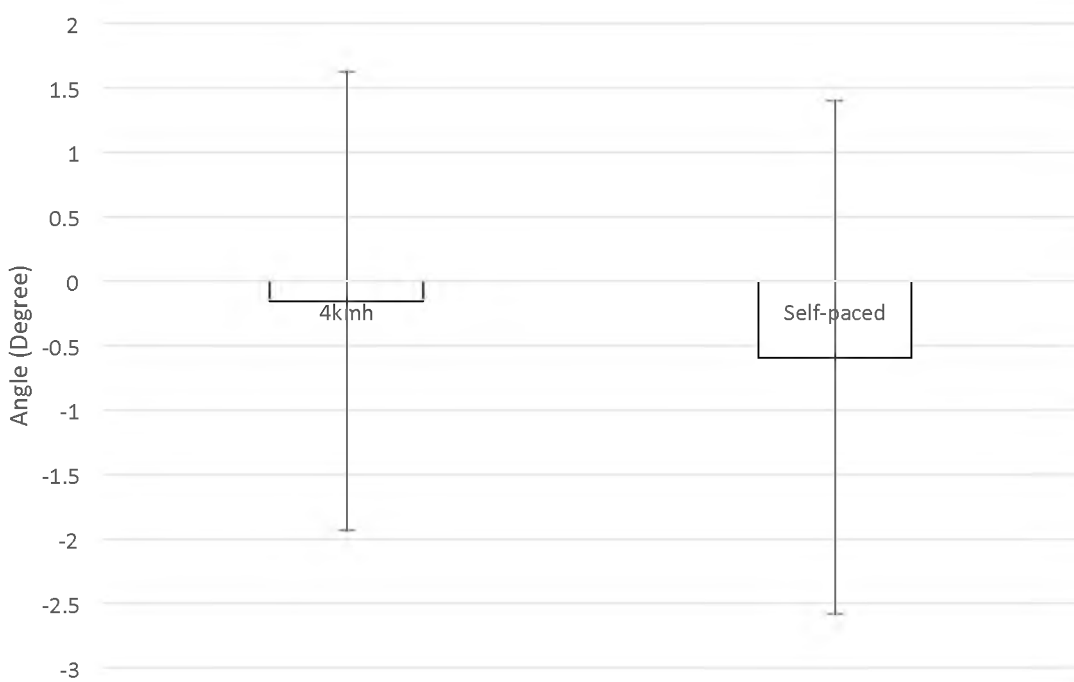

Figure 3-12. Thoracic Lateral Bending by Speed ($p=.005$)

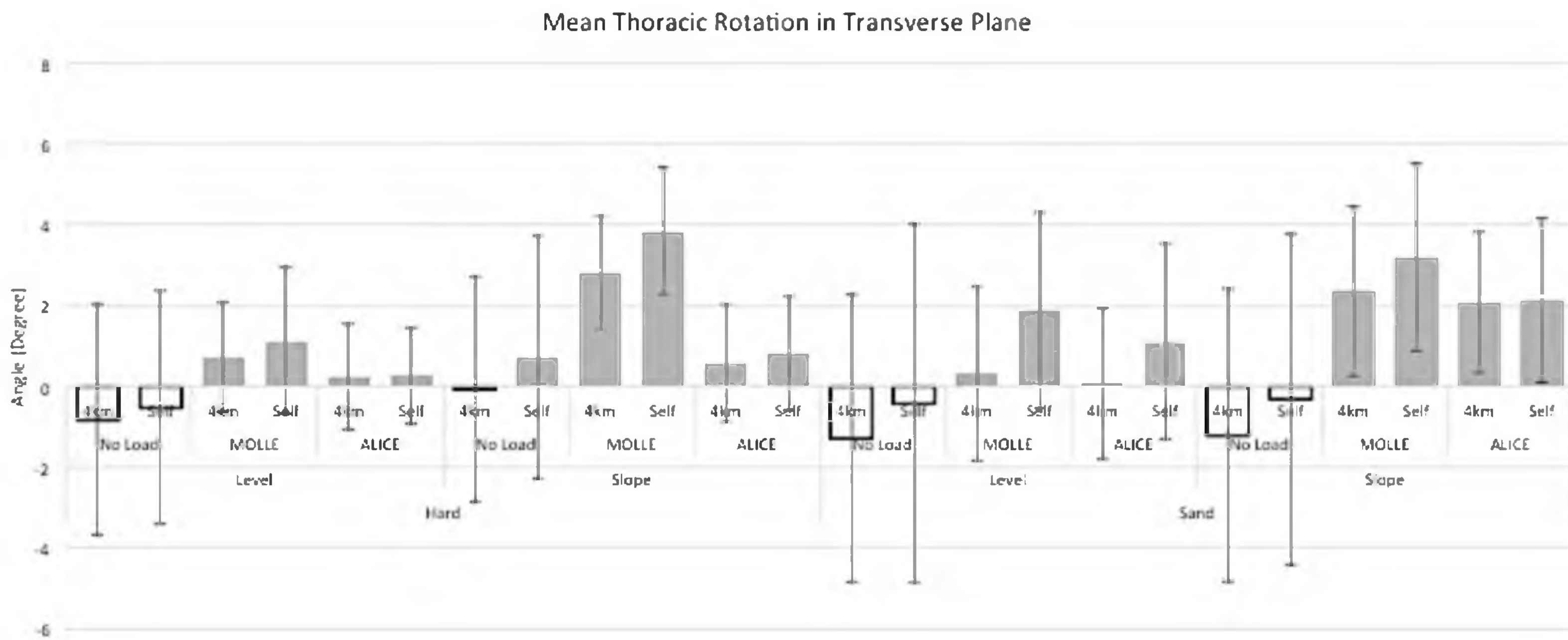

Figure 3-13. Mean Thoracic Rotation in Transverse Plane

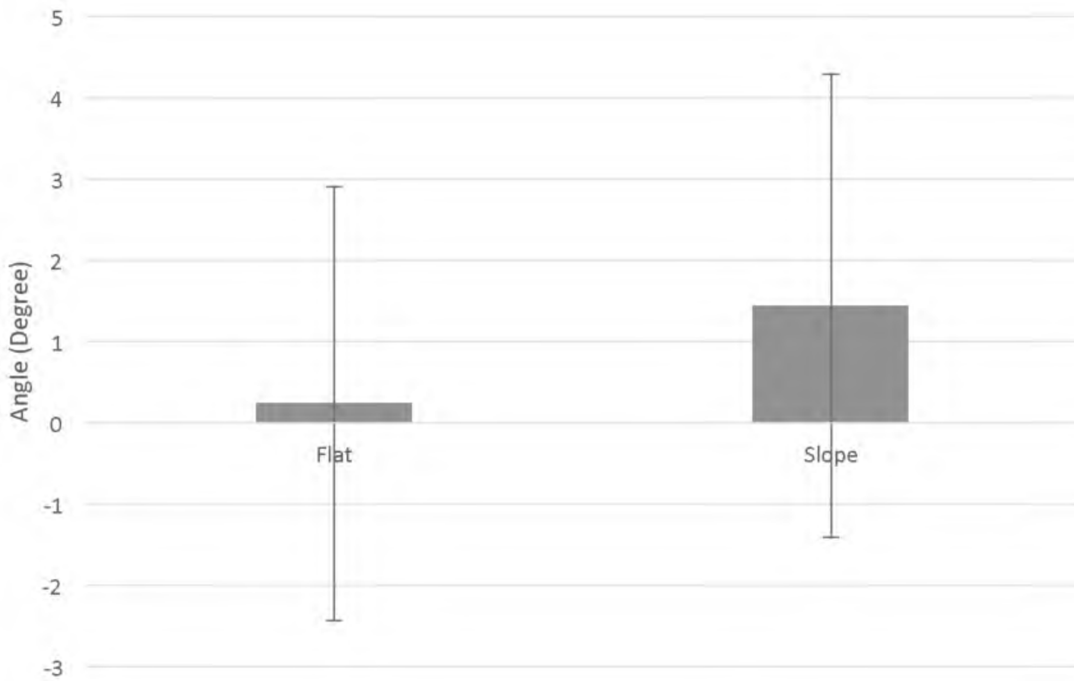

Figure 3-14. Thoracic Rotation by Slope (p=.001)

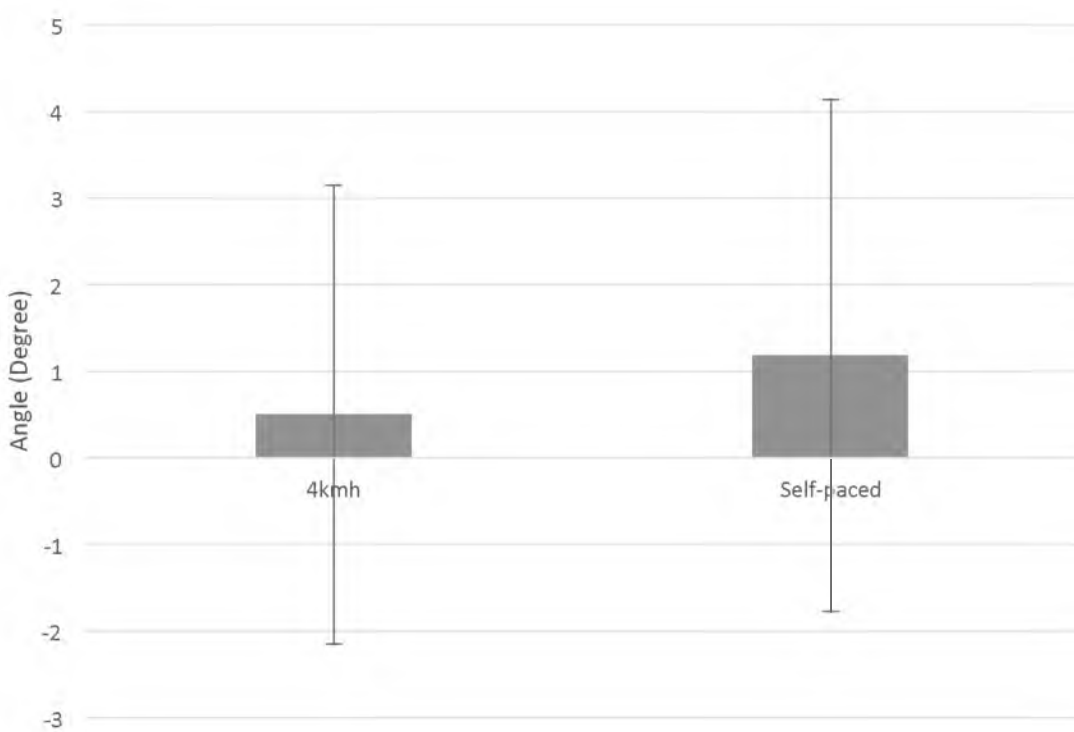

Figure 3-15. Thoracic Rotation by Speed (p=.001)

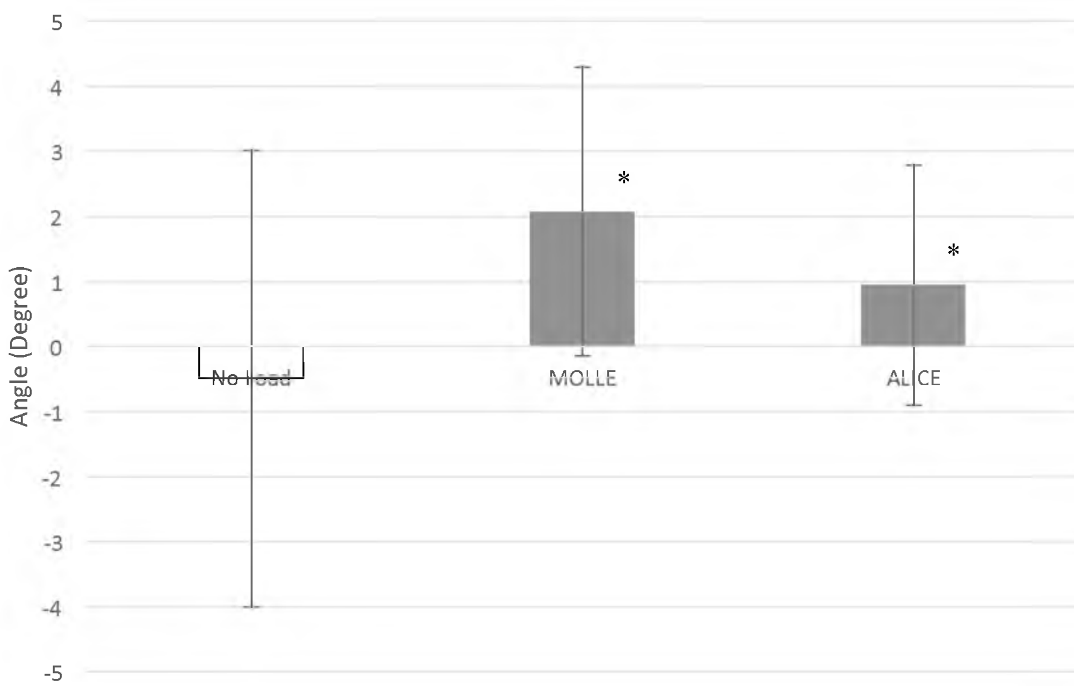

Figure 3-16. Thoracic Rotation by Backpack (p=.001)
* indicates statistical significance (p<.005) compared to No Load

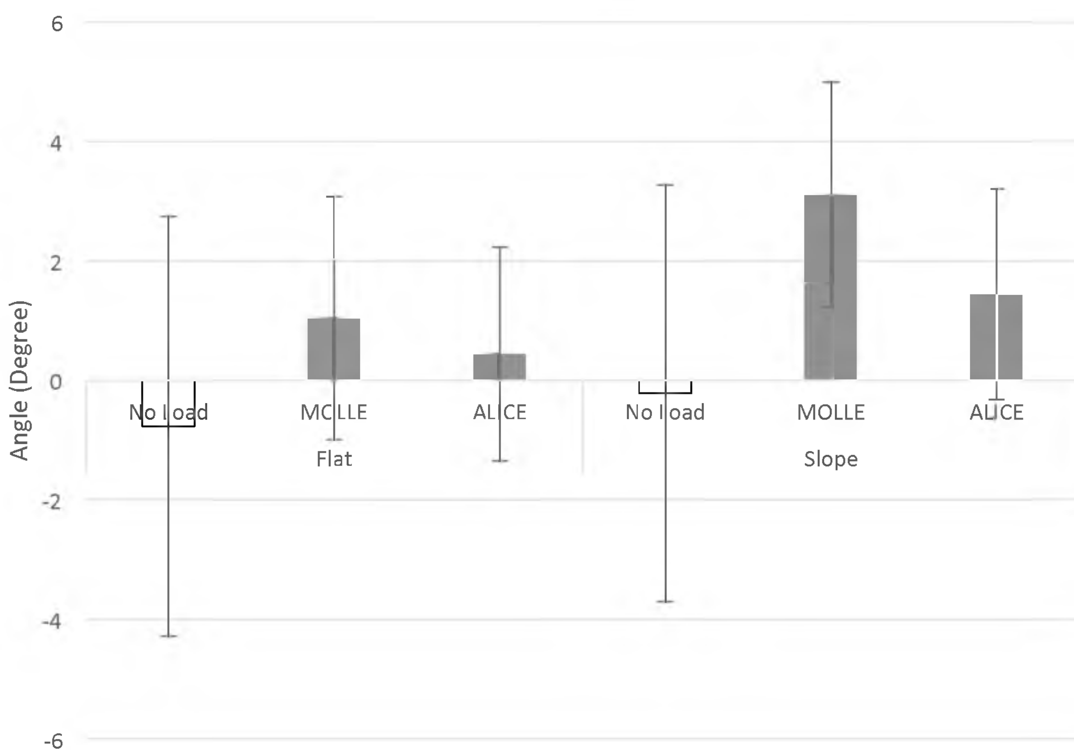

Figure 3-17. Thoracic Rotation by Slope and Backpack (p=.039)

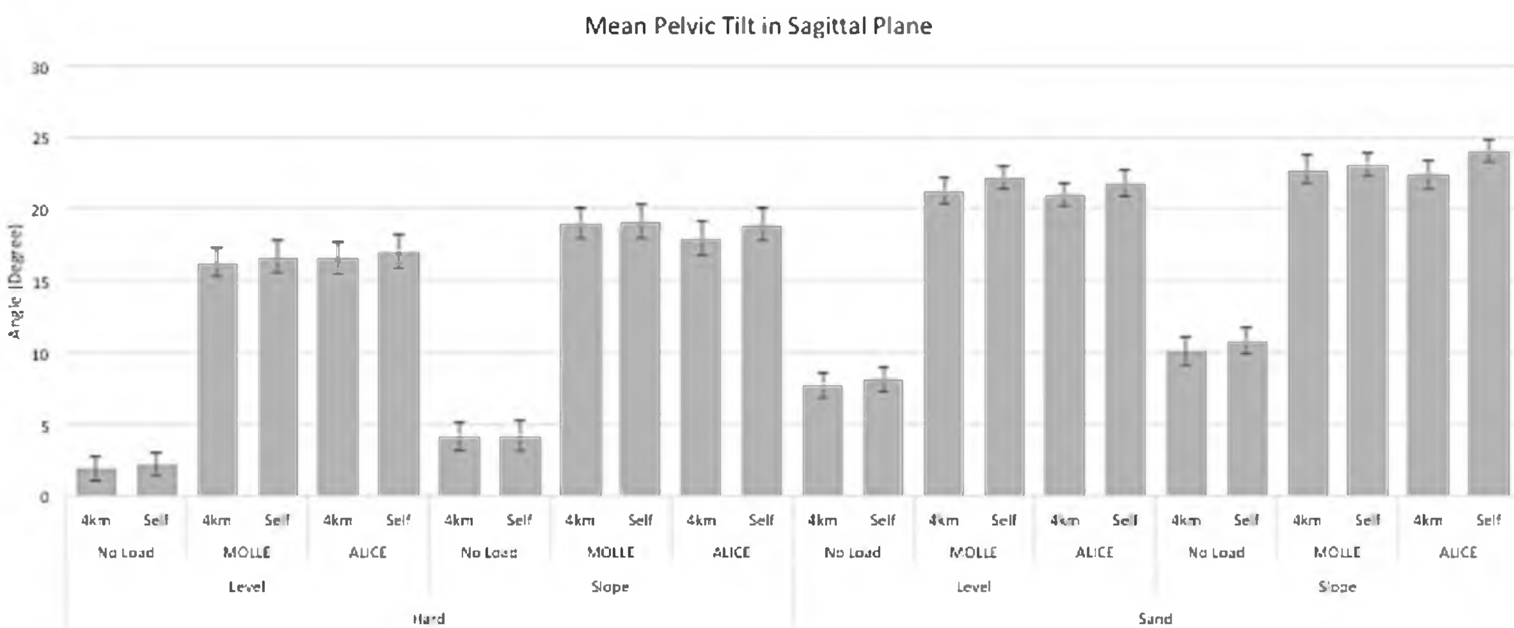

Figure 3-18. Mean Pelvic Tilt in Sagittal Plane

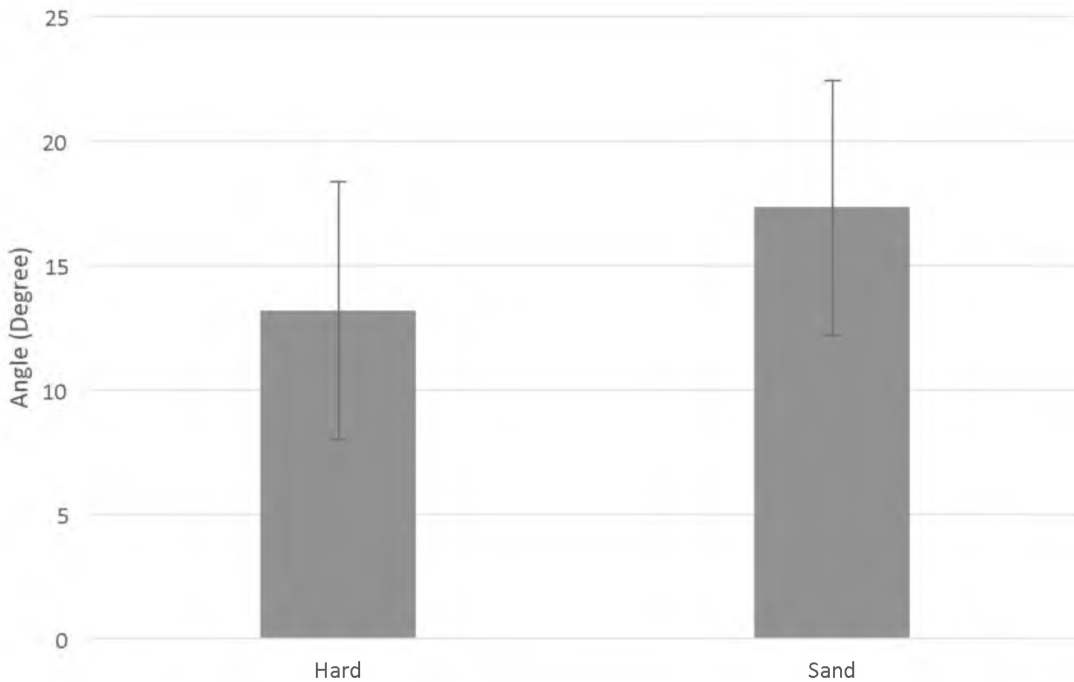

Figure 3-19. Pelvic Tilt by Surface (p<.001)

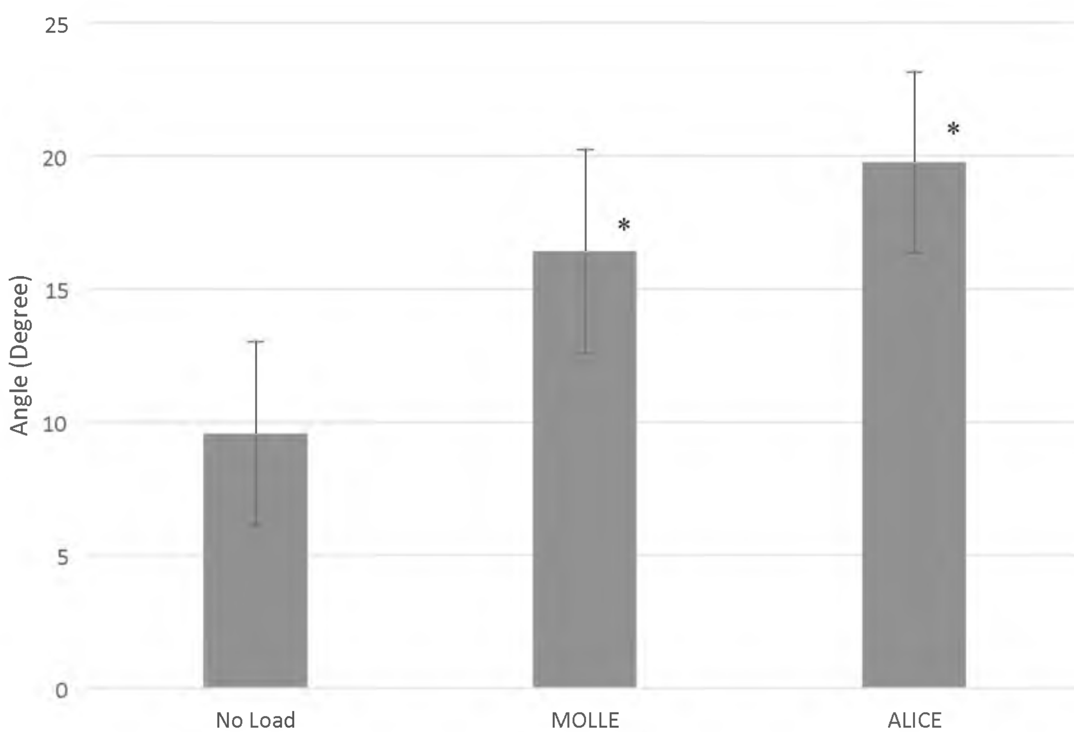

Figure 3-20. Pelvic Tilt by Backpack (p<.001)
* indicates statistical significance (p<.005) compared to No Load

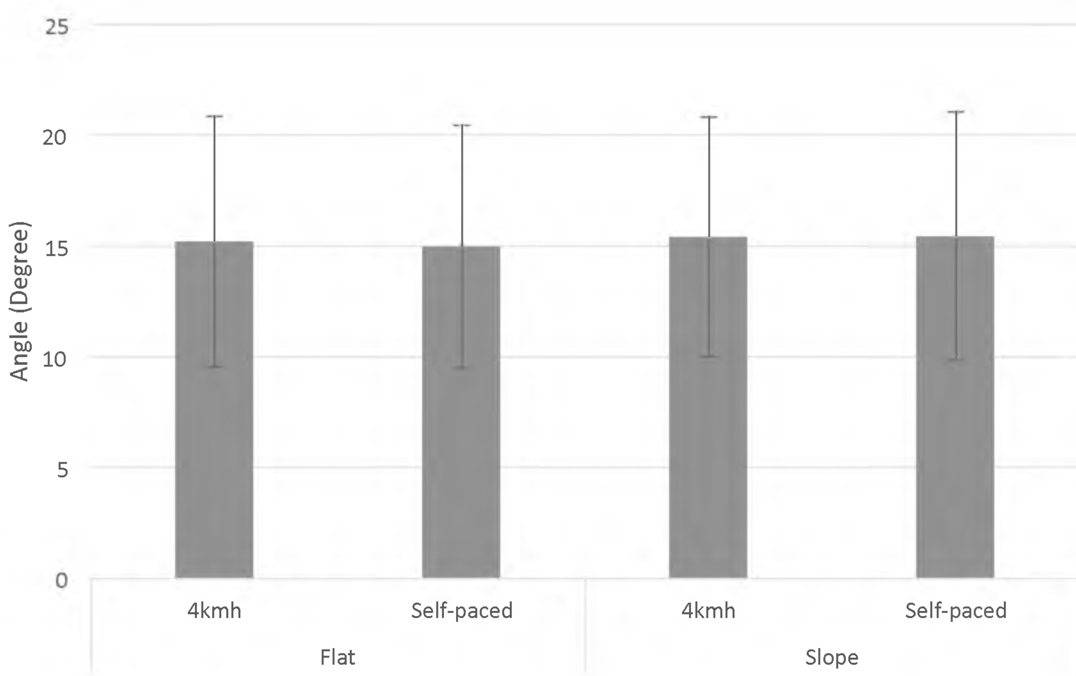

Figure 3-21. Pelvic Tilt by Slope and Speed (*p*=.025)

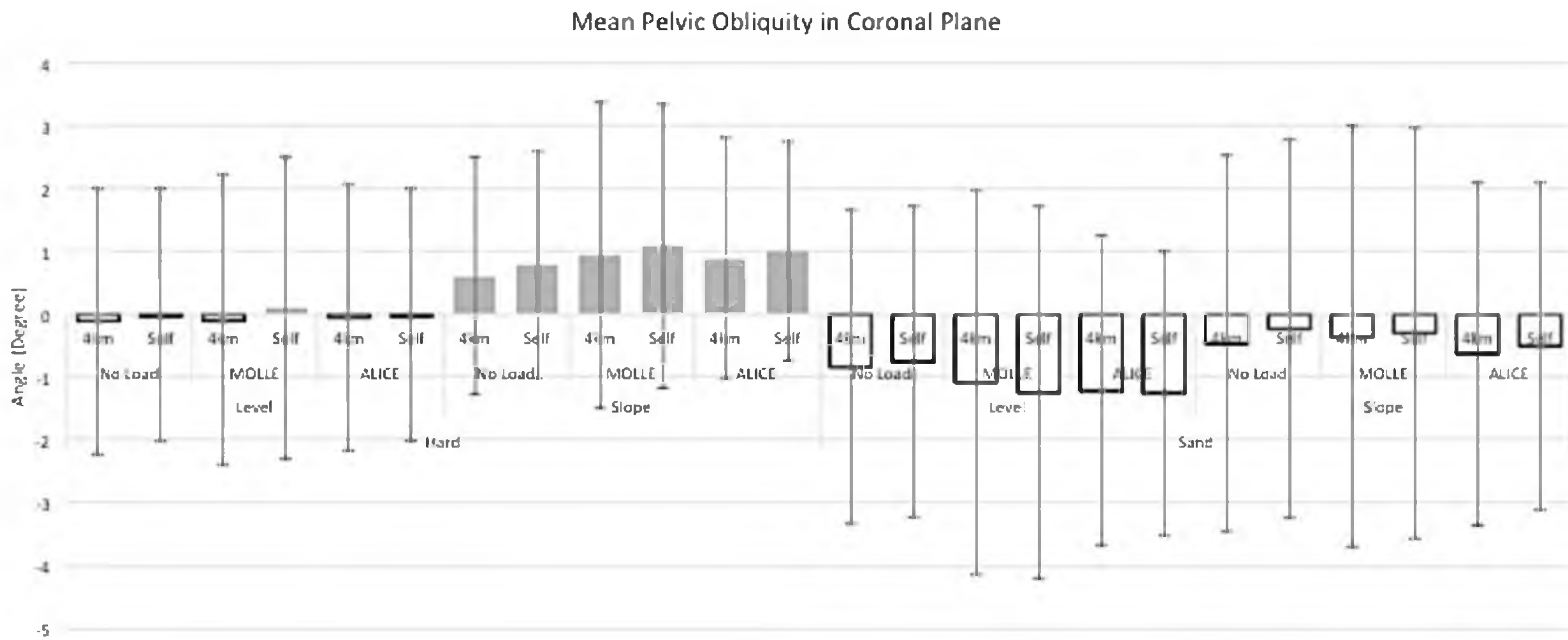

Figure 3-22. Mean Pelvic Obliquity in Coronal Plane

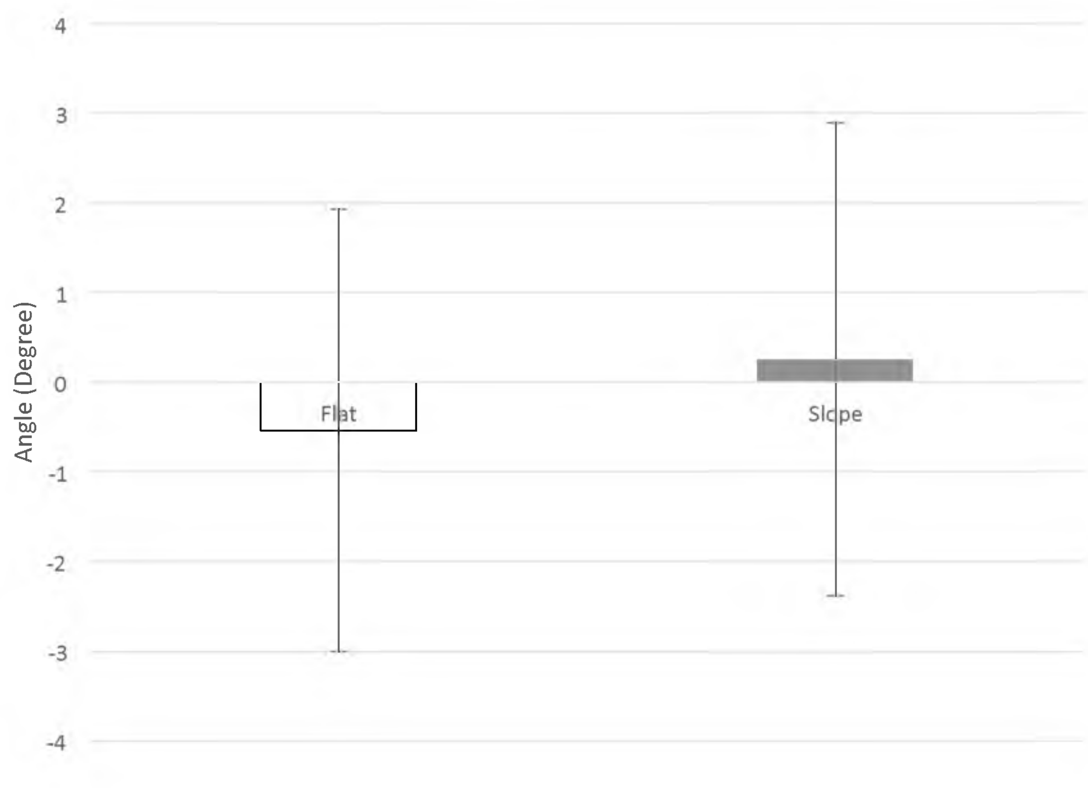

Figure 3-24. Pelvic Obliquity by Slope (p<.001)

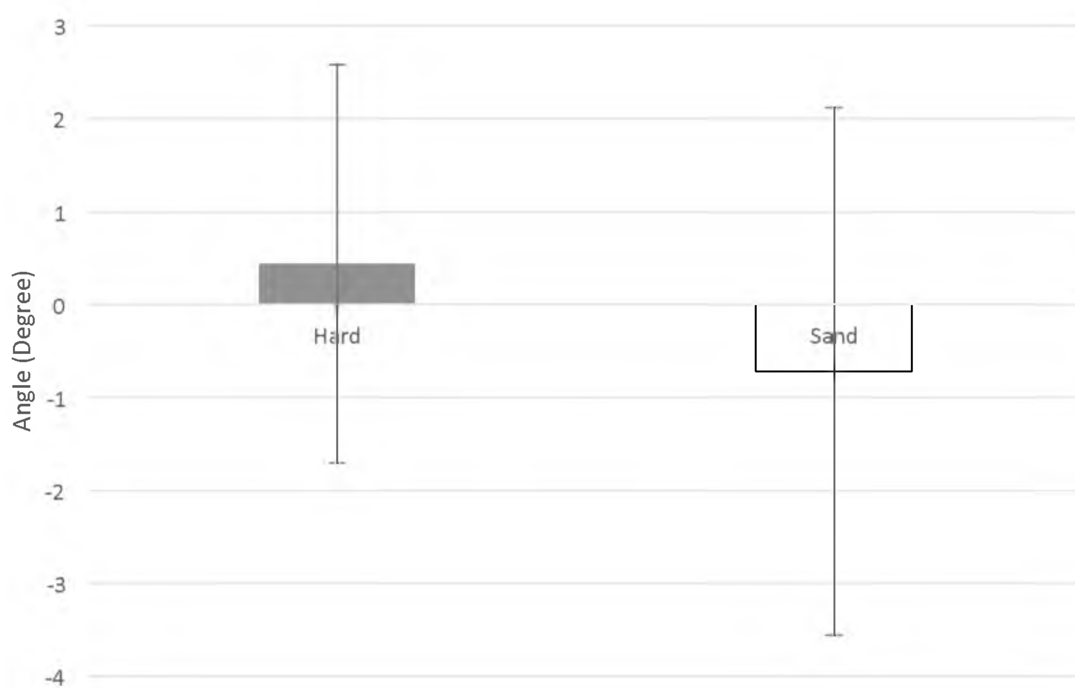

Figure 3-23. Pelvic Obliquity by Surface (p<.001)

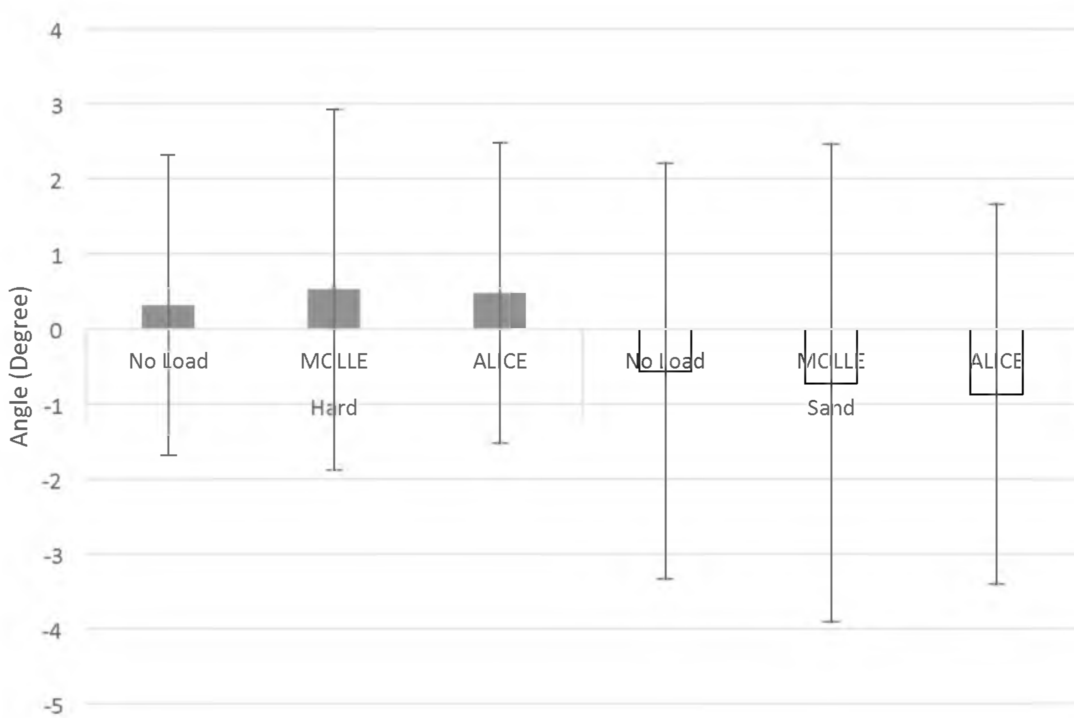

Figure 3-25. Pelvic Obliquity by Surface and Backpack (p=.016)

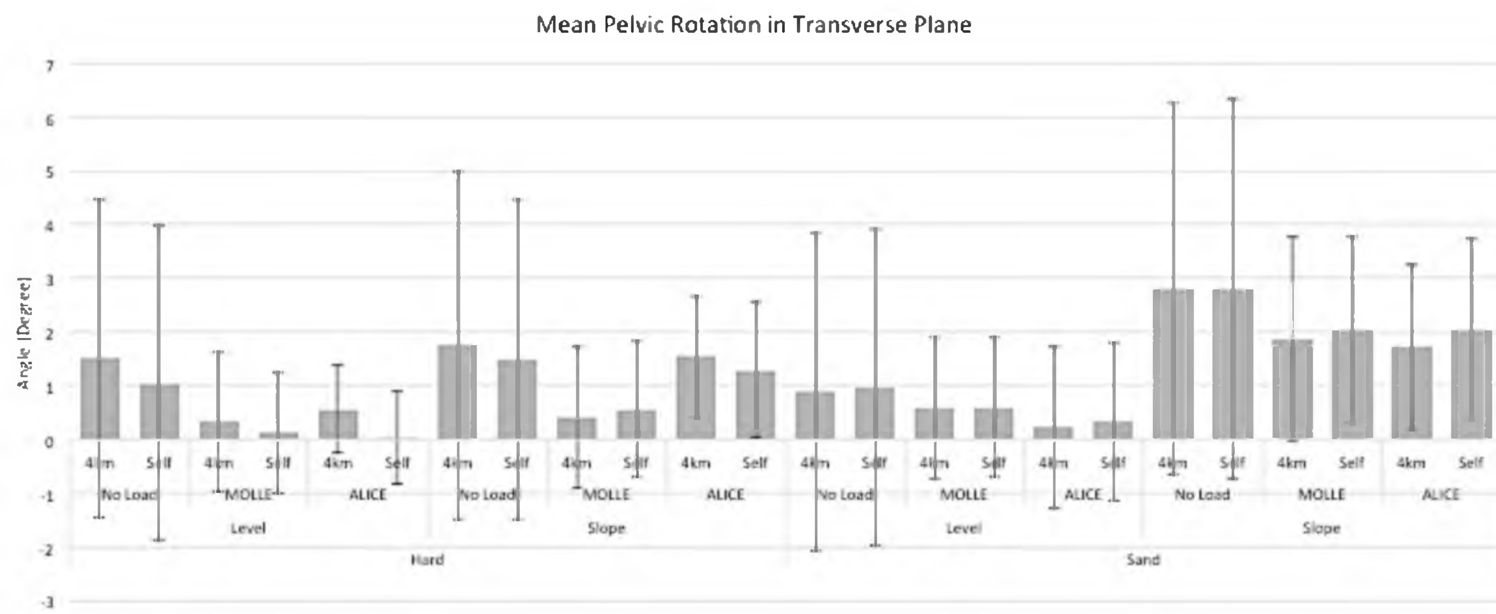

Figure 3-26. Mean Pelvic Rotation in Transverse Plane

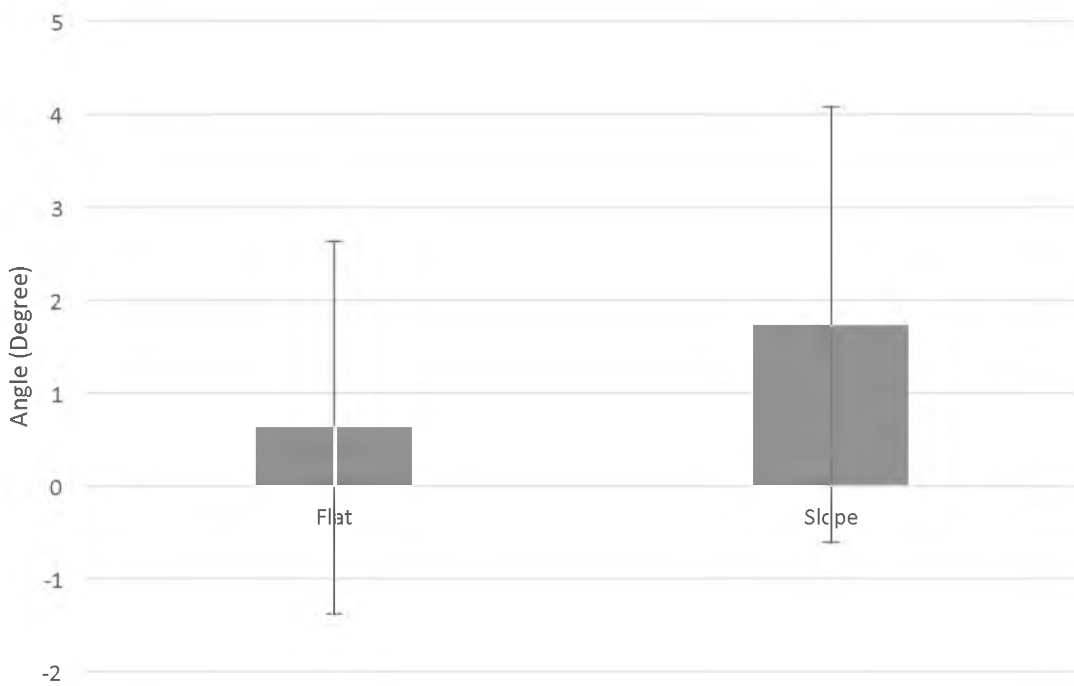

Figure 3-27. Pelvic Rotation by Slope ($p<.001$)

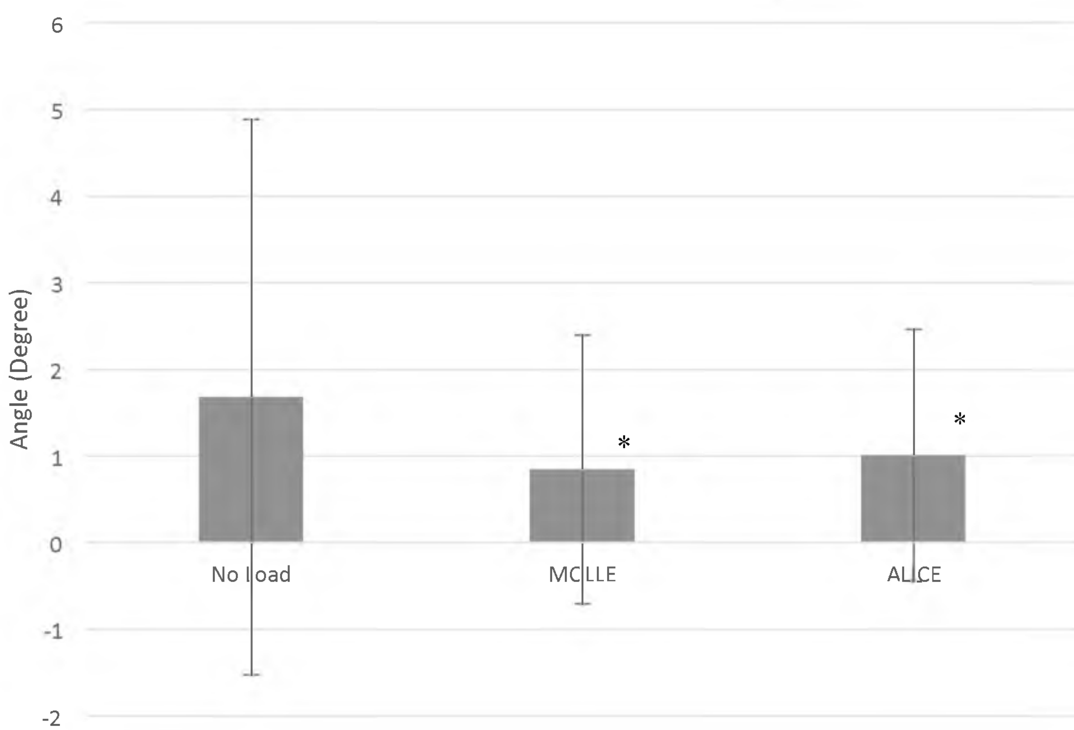

Figure 3-28. Pelvic Rotation by Backpack ($p=.004$)
* indicates statistical significance ($p<.005$) compared to No Load

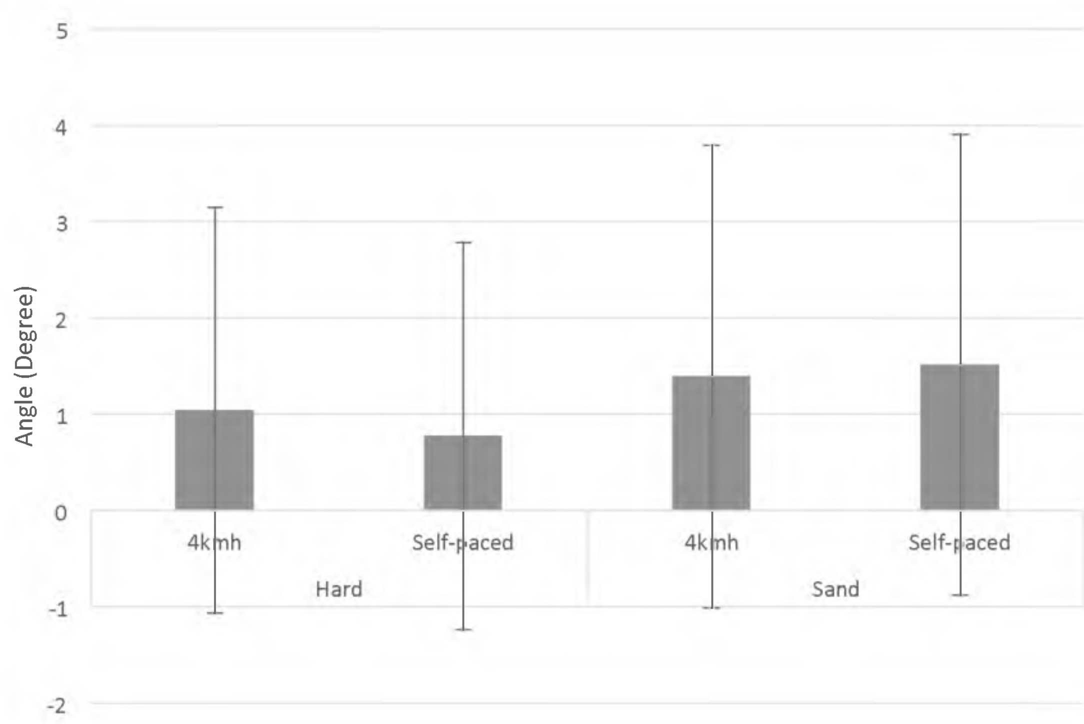

Figure 3-29. Pelvic Rotation by Surface and Speed (p=.012)

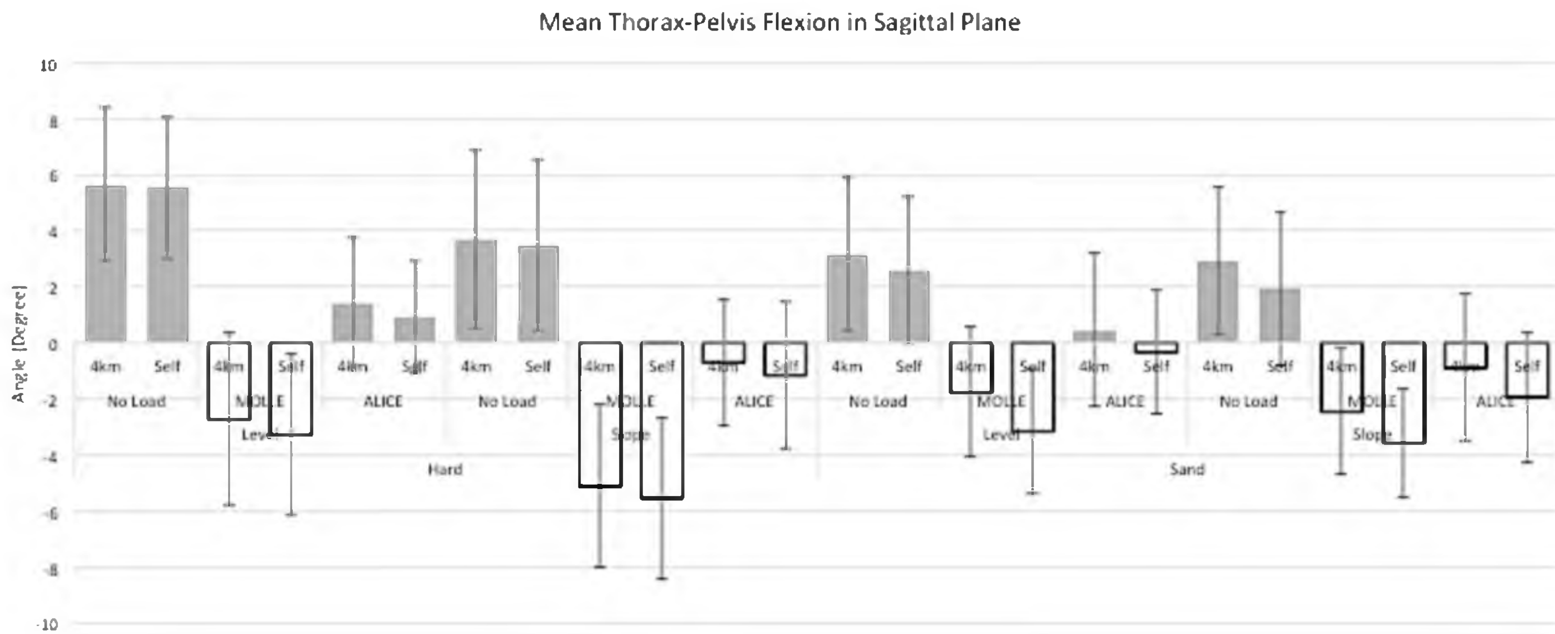

Figure 3-30. Mean Thorax-Pelvis Flexion in Sagittal Plane

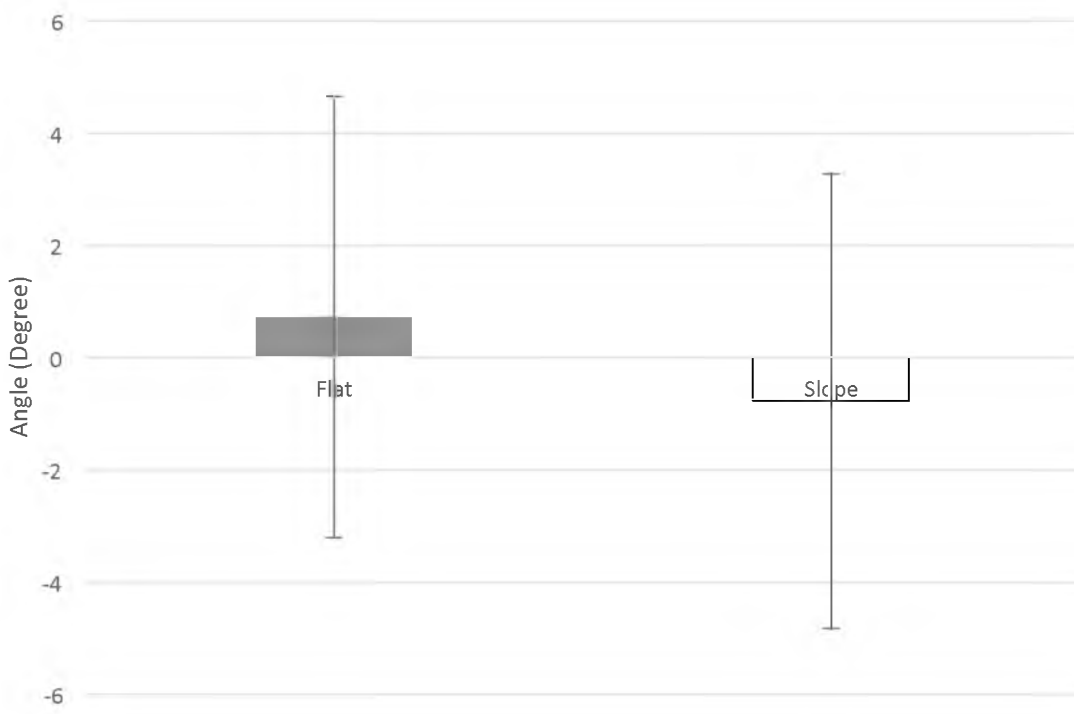

Figure 3-31. Thorax-Pelvis Flexion by Slope (p<.001)

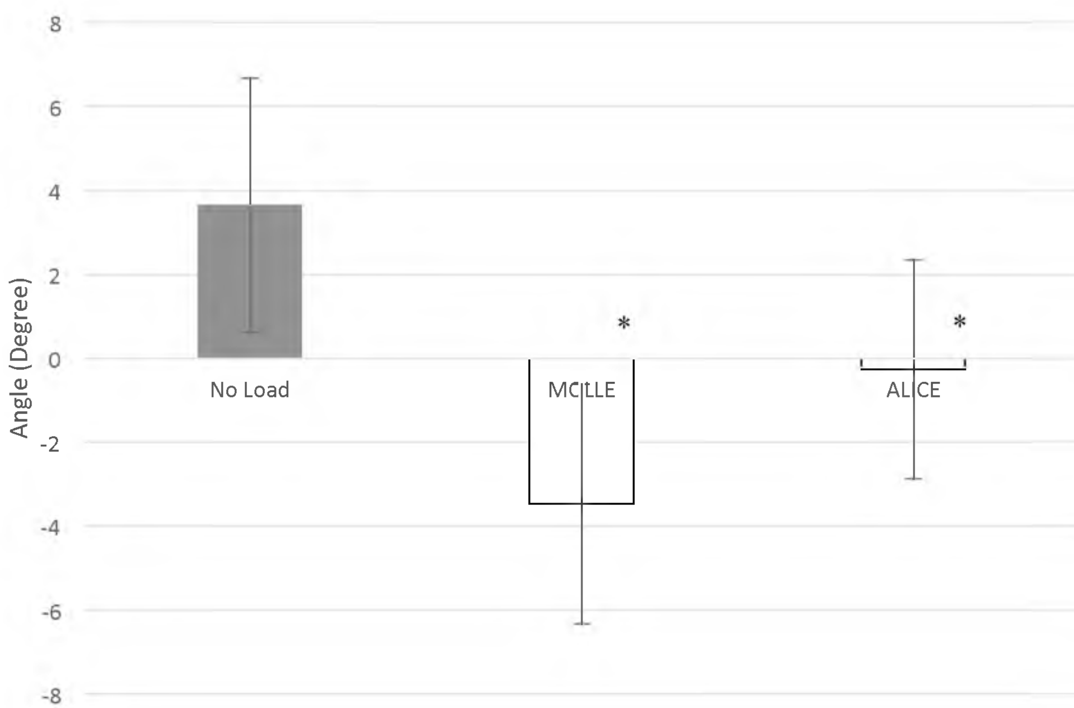

Figure 3-32. Thorax-Pelvis Flexion by Backpack (p<.001)
* indicates statistical significance (p<.005) compared to No Load

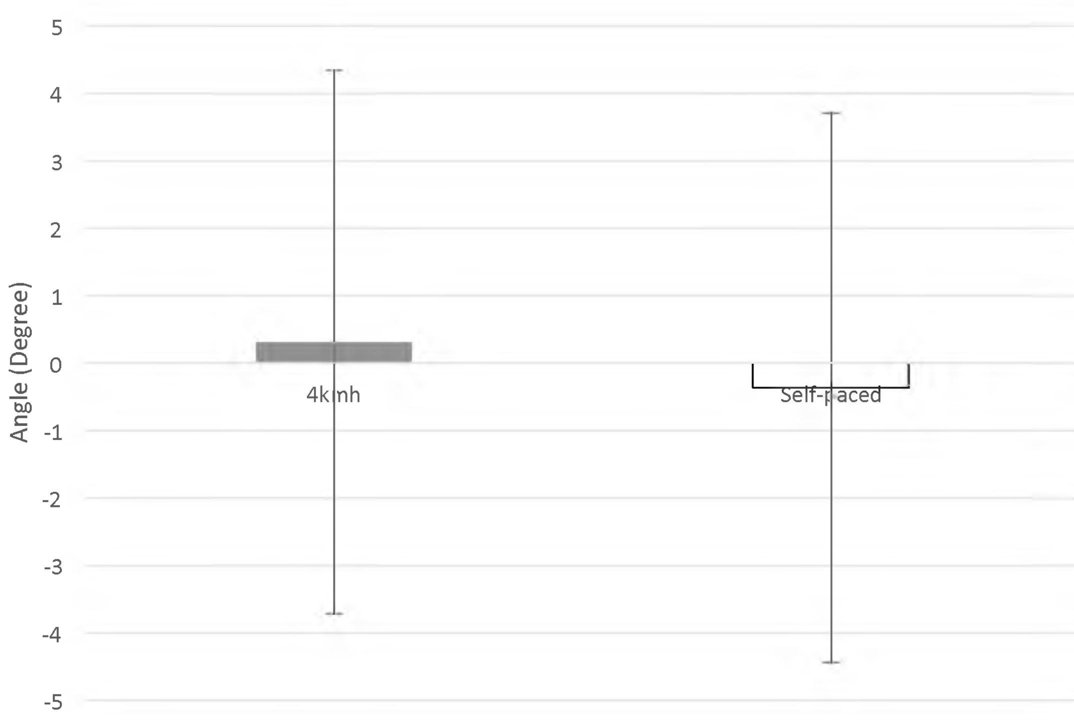

Figure 3-33. Thorax-Pelvis Flexion by Speed (p=.001)

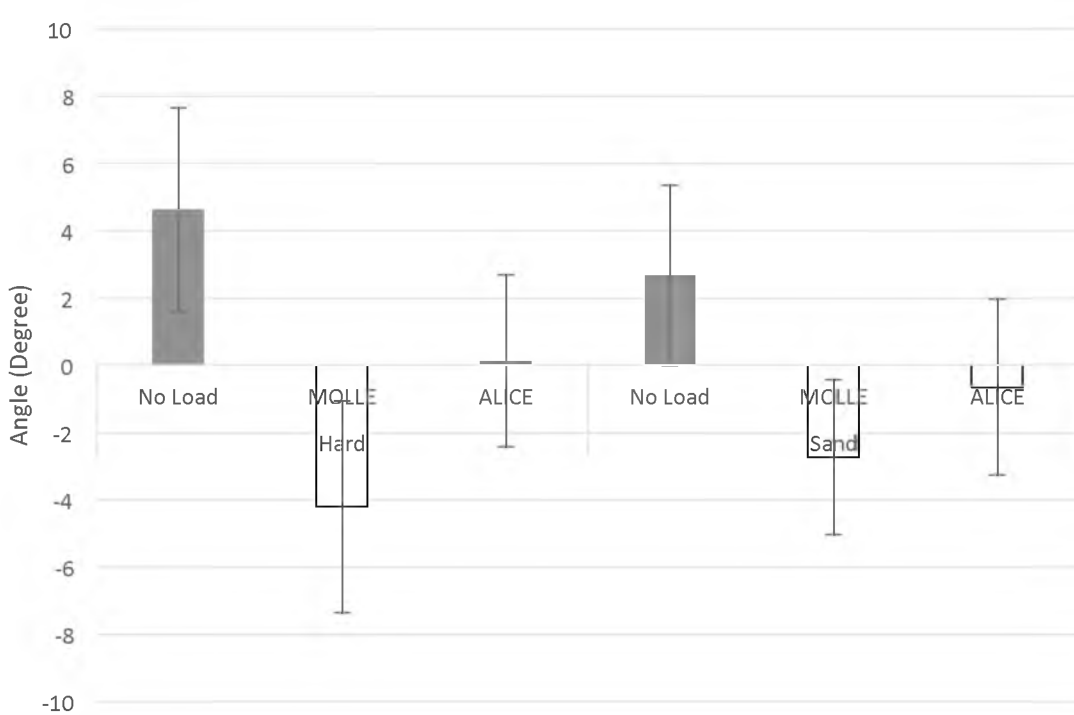

Figure 3-34. Thorax-Pelvis Flexion by Surface and Backpack (p=.009)

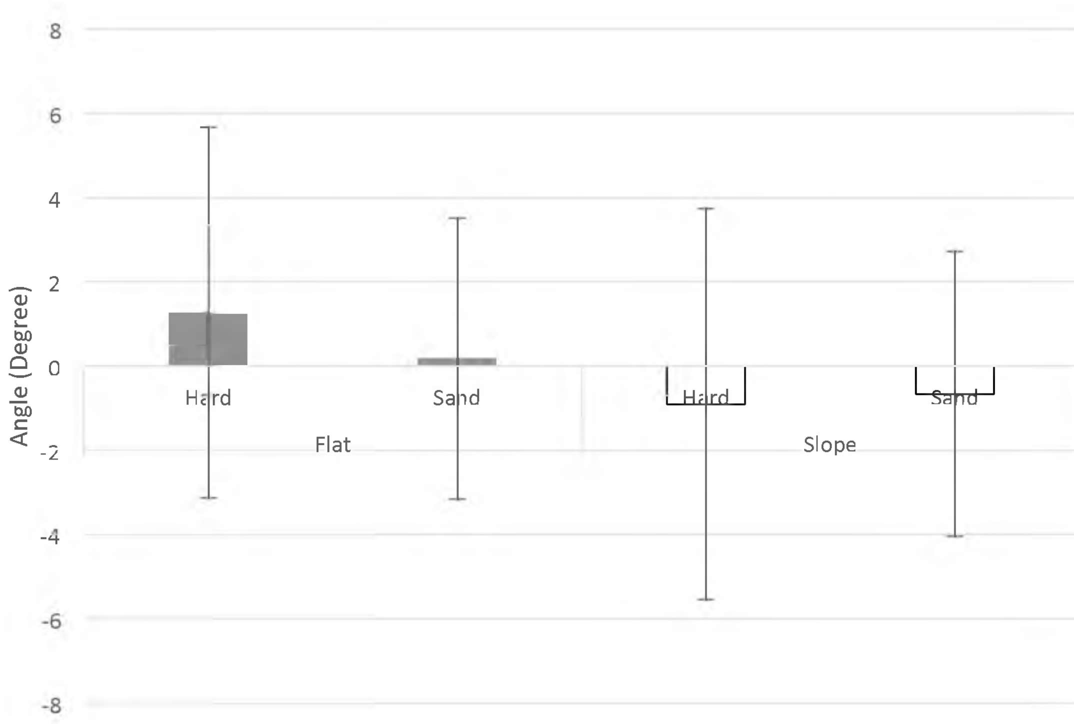

Figure 3-35. Thorax-Pelvis Flexion by Slope and Surface (p<.001)

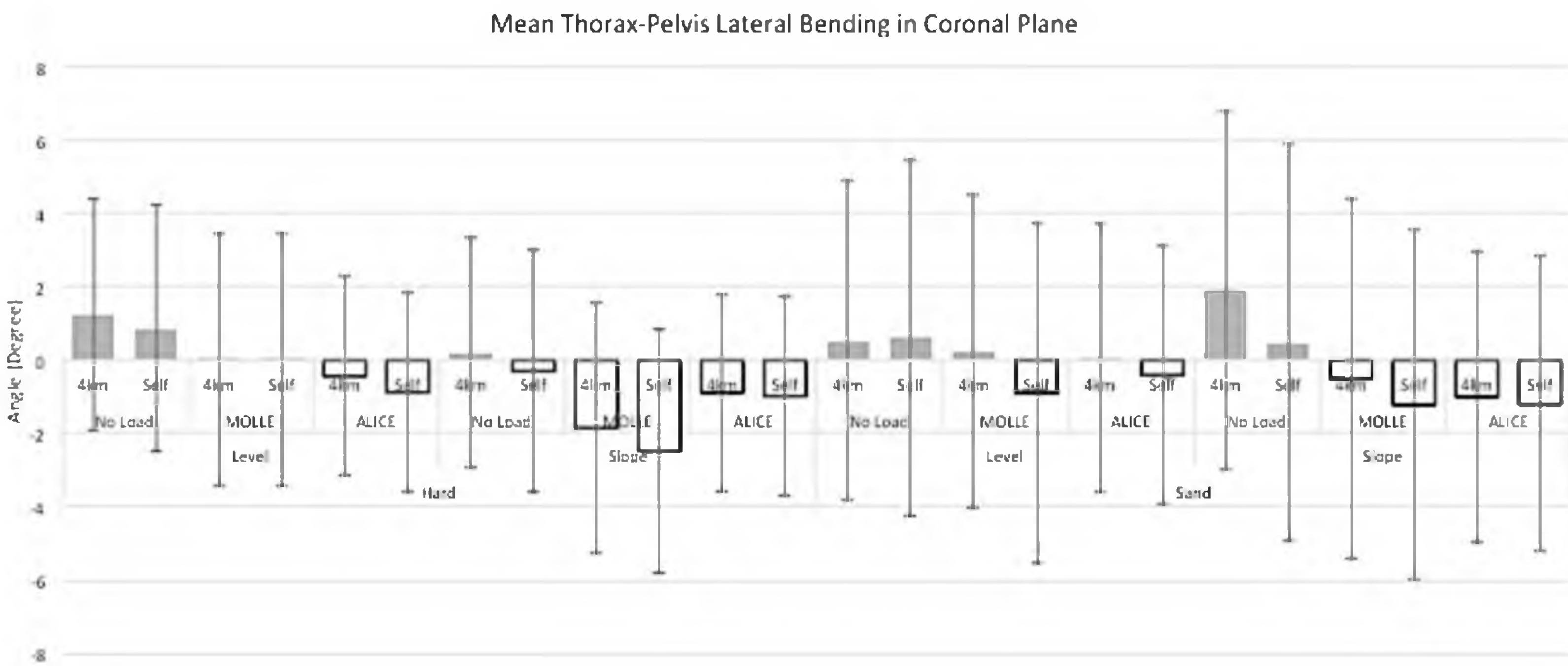

Figure 3-36. Mean Thorax-Pelvis Lateral Bending in Coronal Plane

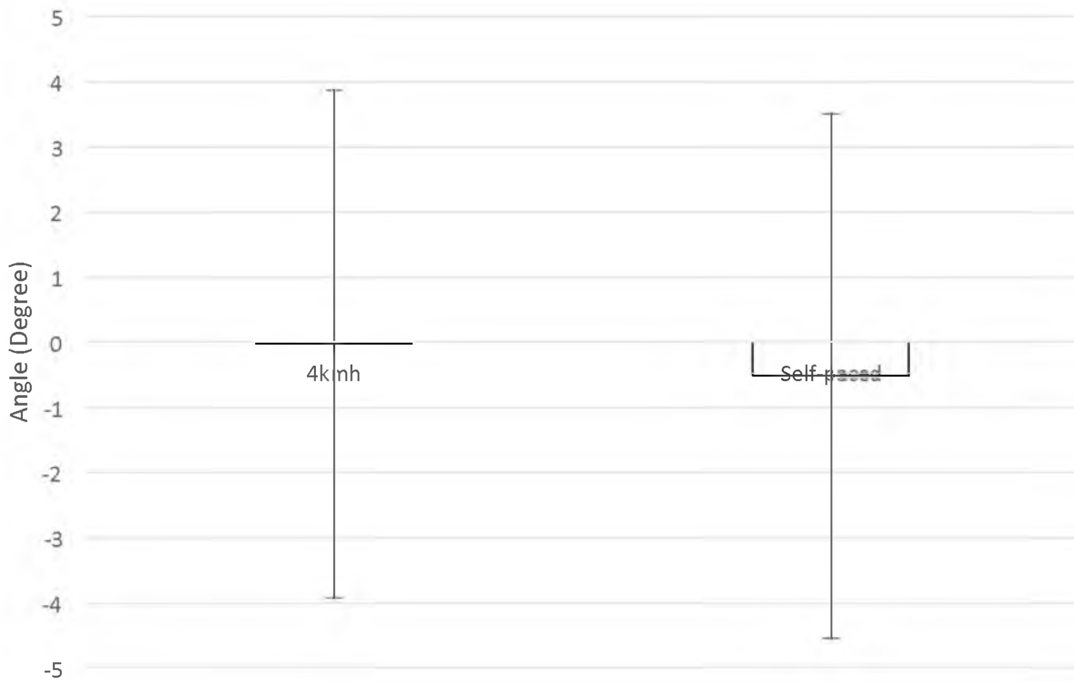

Figure 3-37. Thorax-Pelvis Lateral Bending by Speed (p=.034)

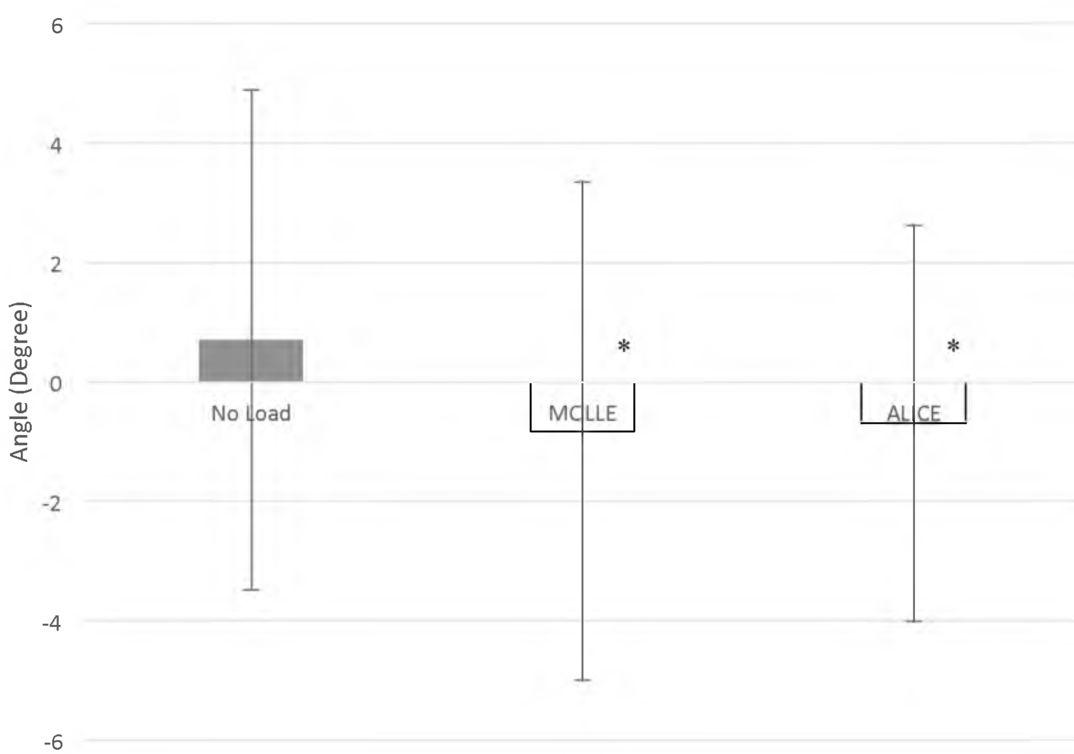

Figure 3-38. Thorax-Pelvis Lateral Bending by Backpack (p<.001)
* indicates statistical significance (p<.005) compared to No Load

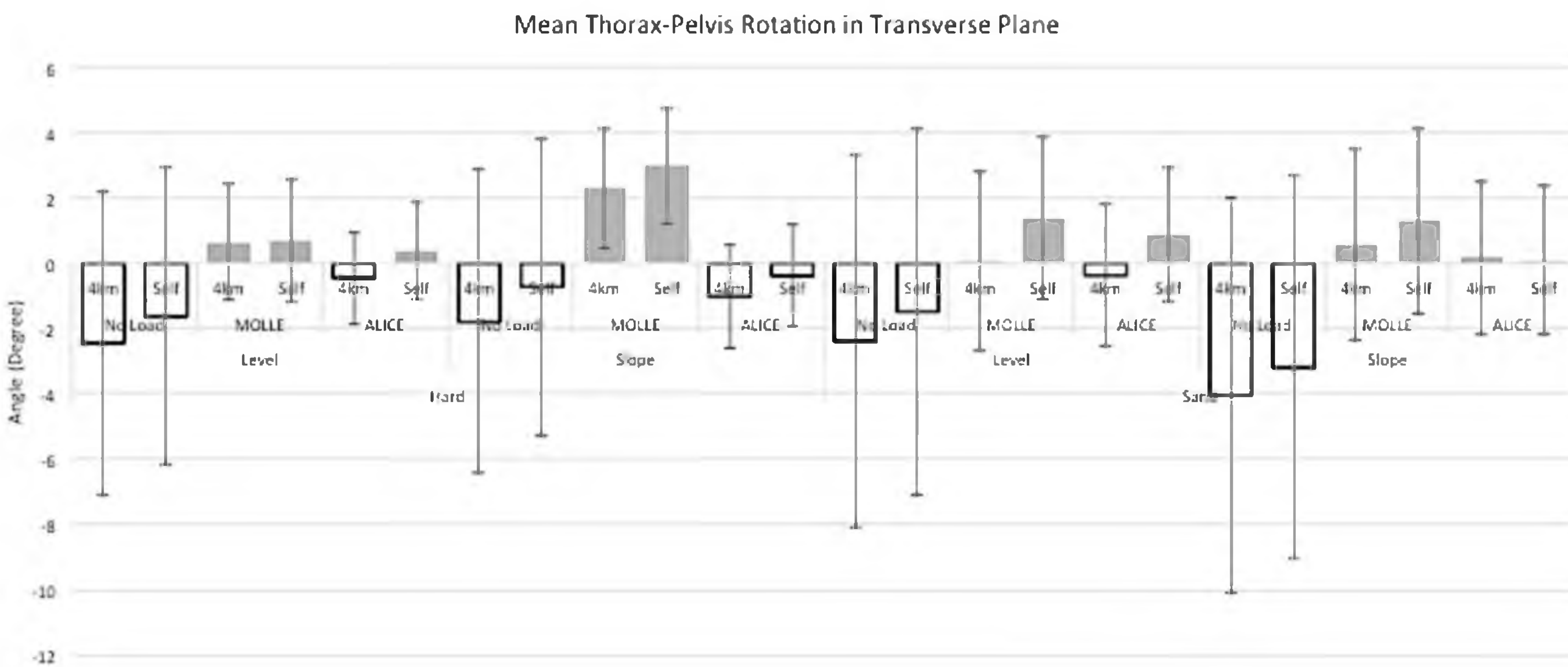

Figure 3-39. Mean Thorax-Pelvis Rotation in Transverse Plane

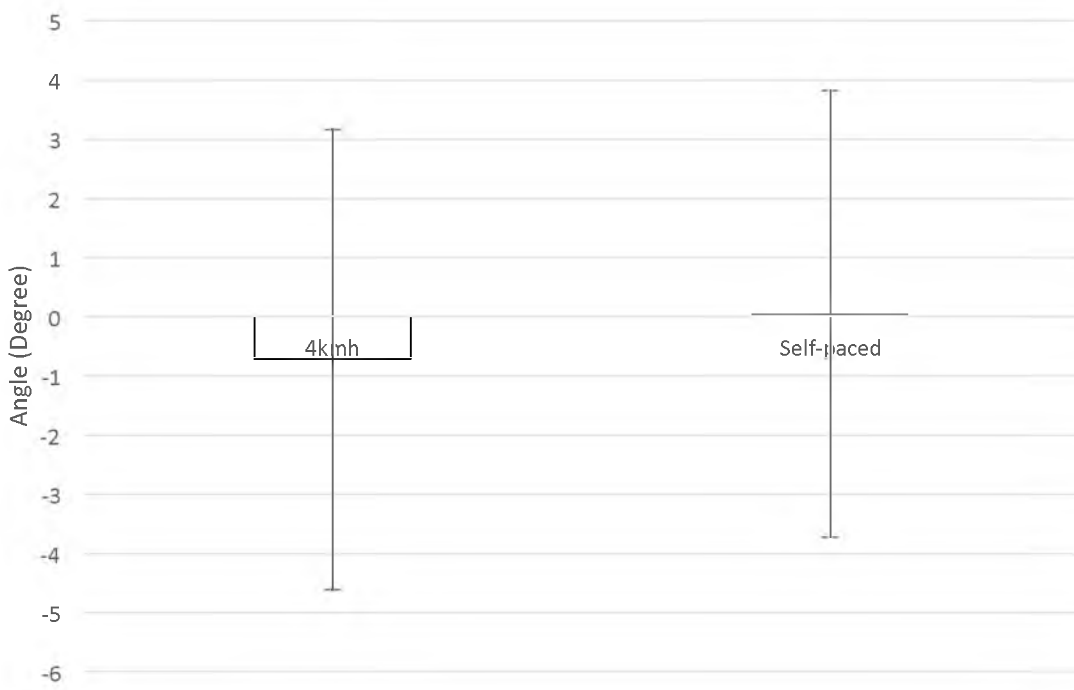

Figure 3-40. Thorax-Pelvis Rotation by Speed ($p<.001$)

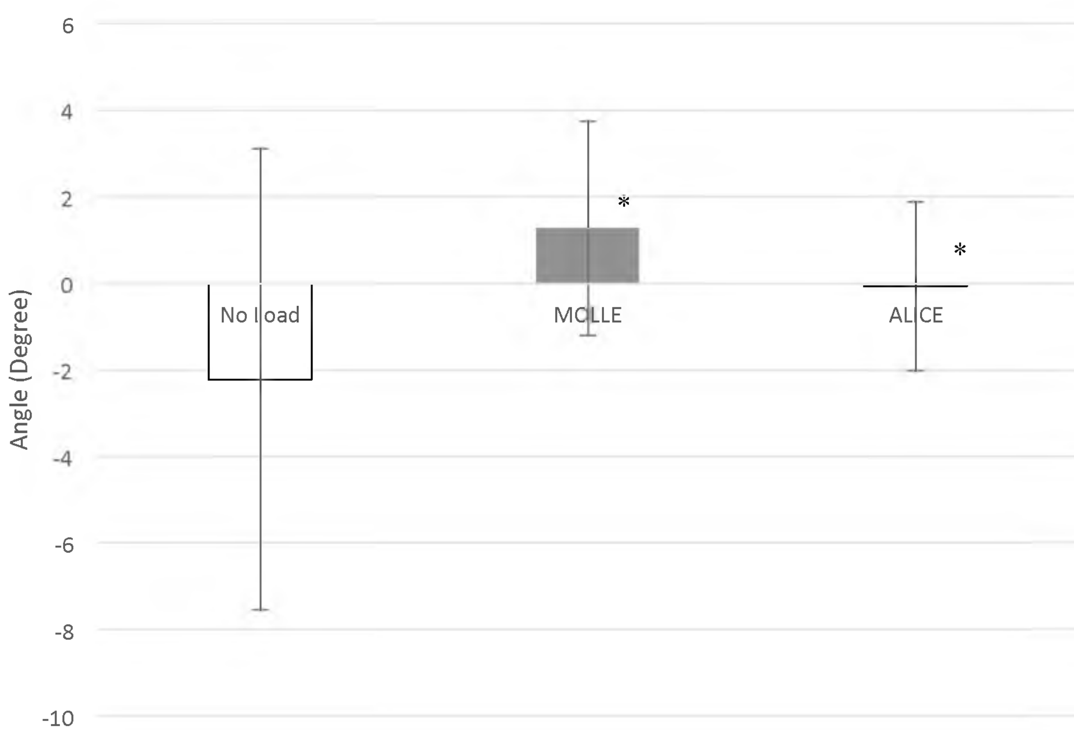

Figure 3-41. Thorax-Pelvis Rotation by Backpack ($p<.001$)
* indicates statistical significance ($p<.005$) compared to No Load

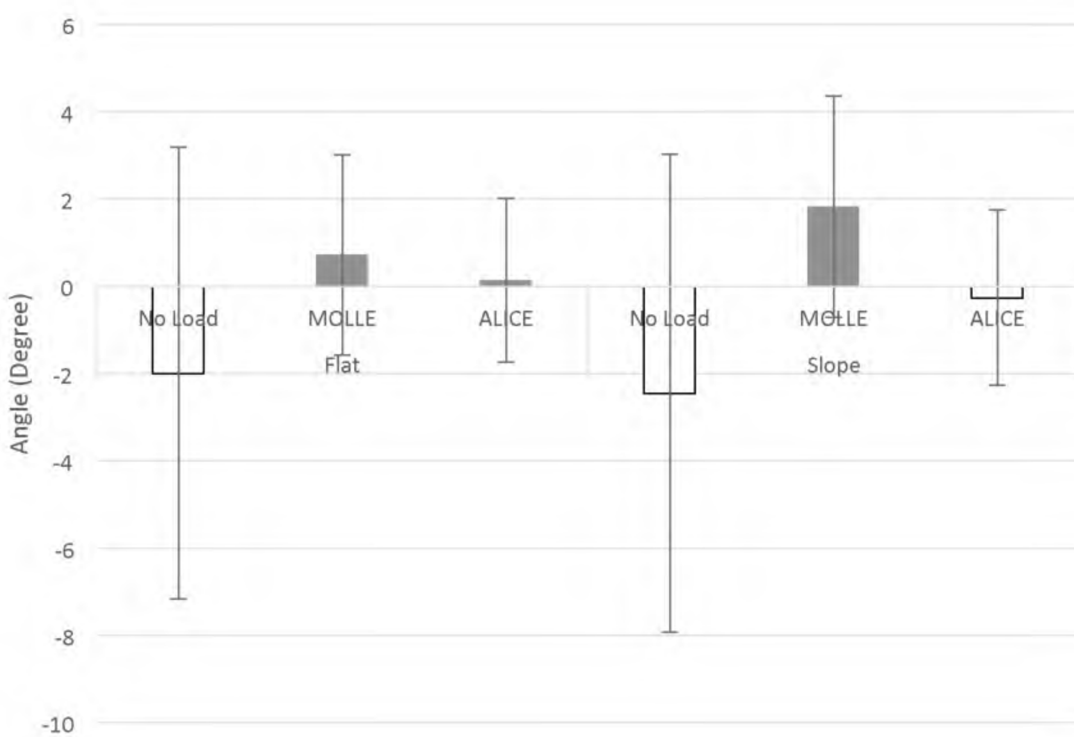

Figure 3-42. Thorax-Pelvis Rotation by Slope and Backpack (p=.041)

Table 3-1. Effect of Loading Conditions on Kinematics

		Significant Effect	Post hoc		
			No vs. MOLLE	No vs. ALICE	MOLLE vs. ALICE
Thorax	Flex/Ext.	Yes	0.000	0.000	0.840
	Lateral Flexion	Yes	0.000	0.000	0.559
	Rotation	Yes	0.000	0.05	0.001
Pelvis	Tilt	Yes	0.000	0.000	0.000
	Obliquity	No	N/A	N/A	N/A
	Rotation	Yes	0.003	0.022	0.460
Relative motion of thorax to pelvis	Flex/Ext.	Yes	0.000	0.003	0.001
	Lateral Flexion	Yes	0.000	0.008	0.115
	Rotation	Yes	0.000	0.003	0.002

Note: Numbers in the table represent *p* values.

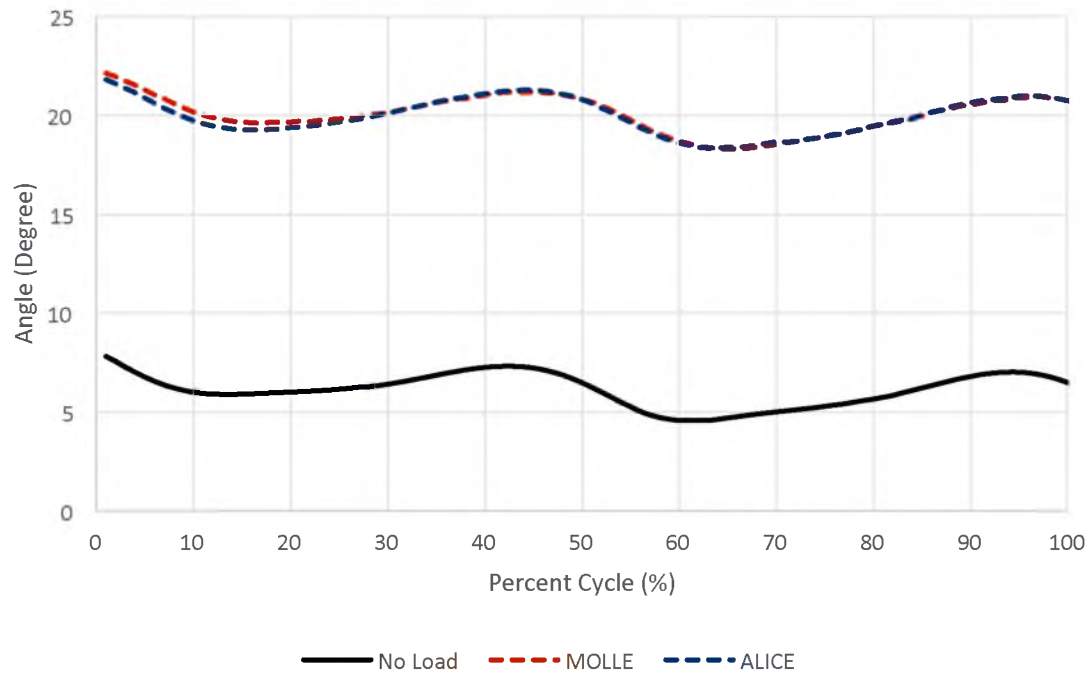

Figure 3-43. Thoracic Flexion Profile (Sagittal)

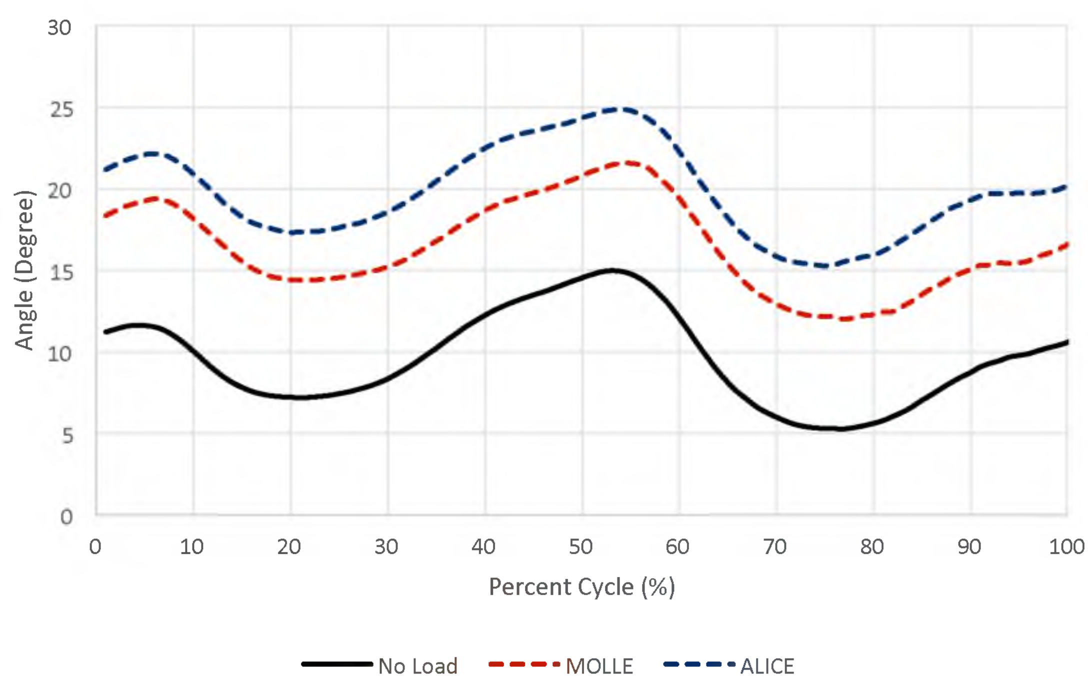

Figure 3-44. Pelvic Tilt Profile (Sagittal)

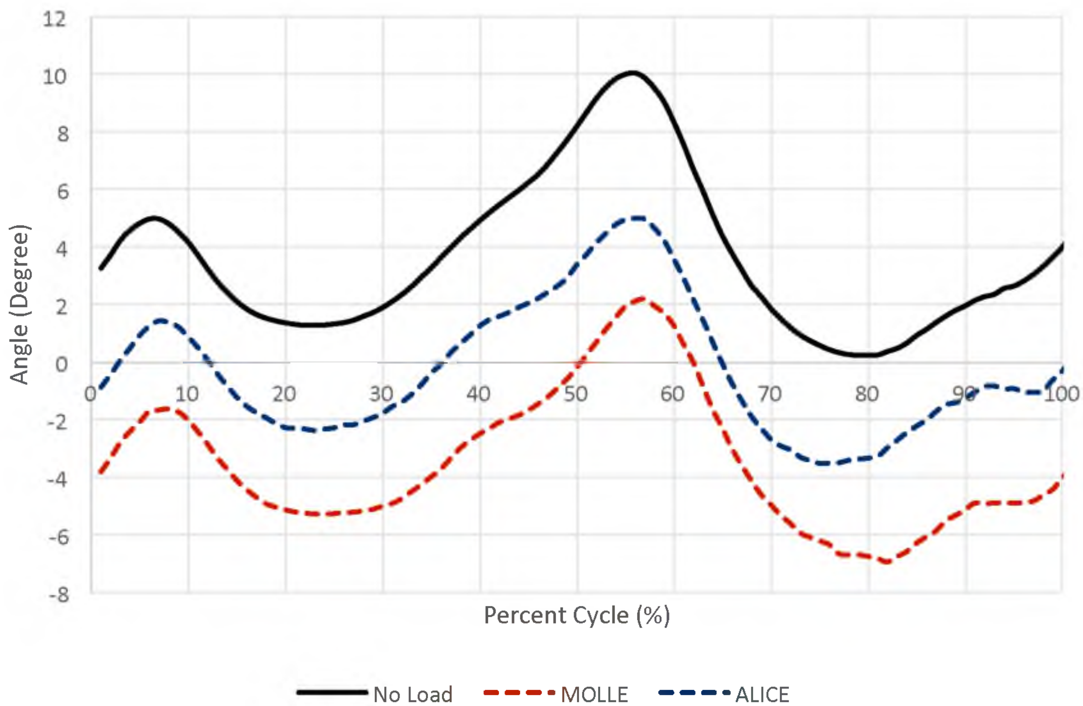

Figure 3-45. Thorax-Pelvis Flexion Profile (Sagittal)

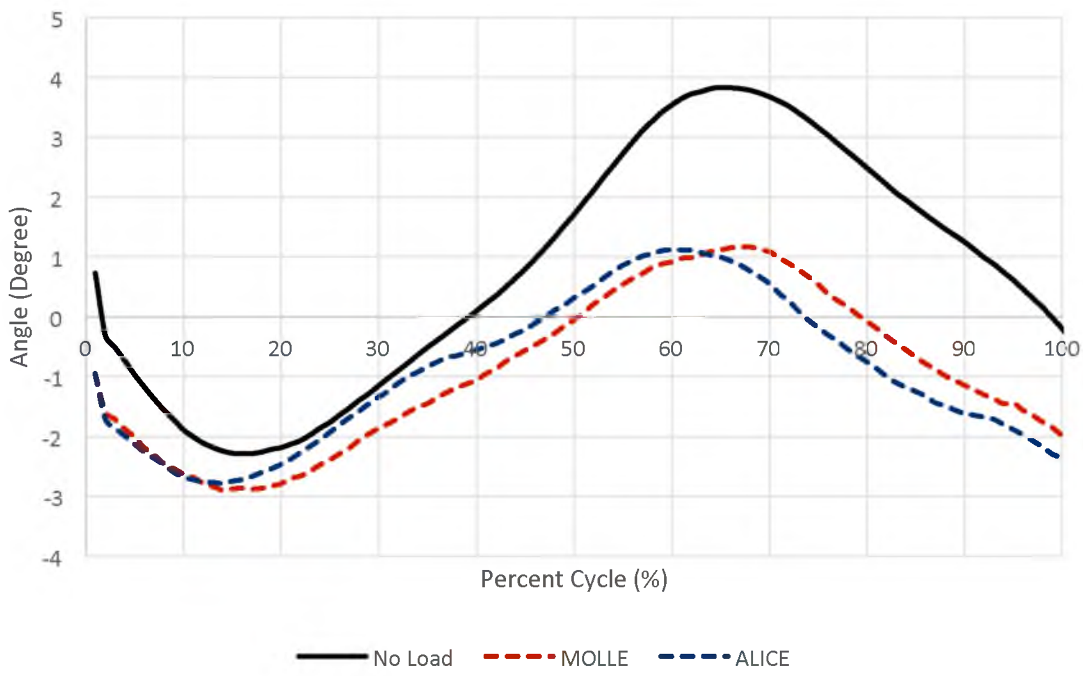

Figure 3-46. Thoracic Lateral Bending Profile (Coronal)

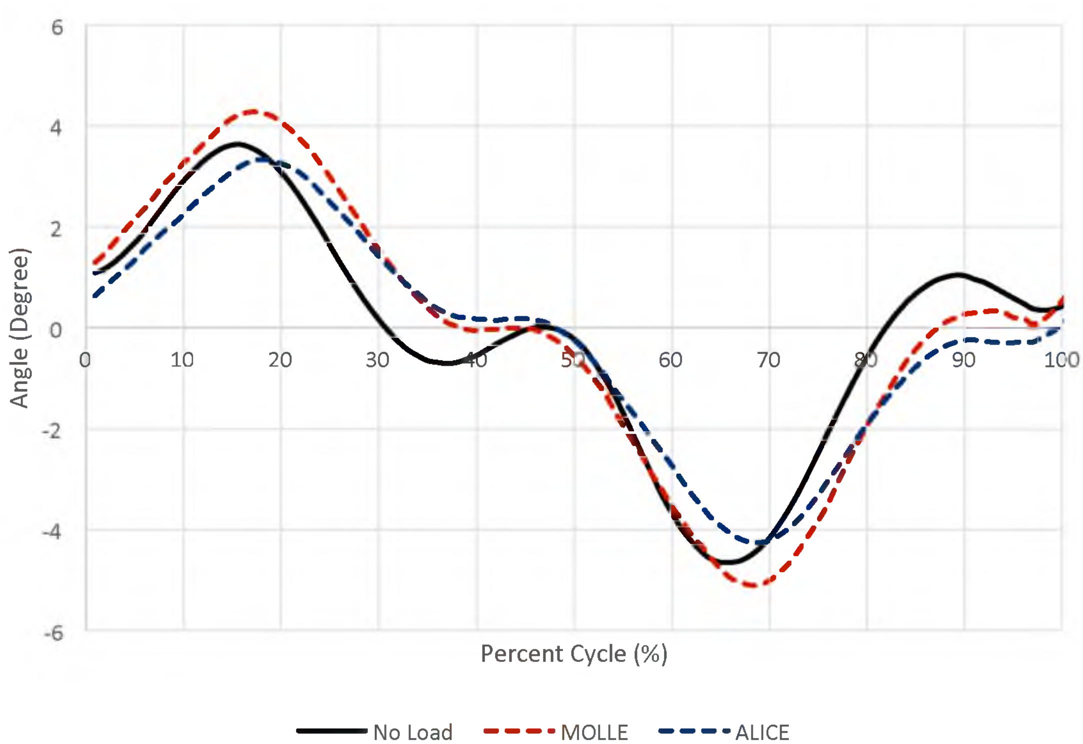

Figure 3-47. Pelvic Obliquity Profile (Coronal)

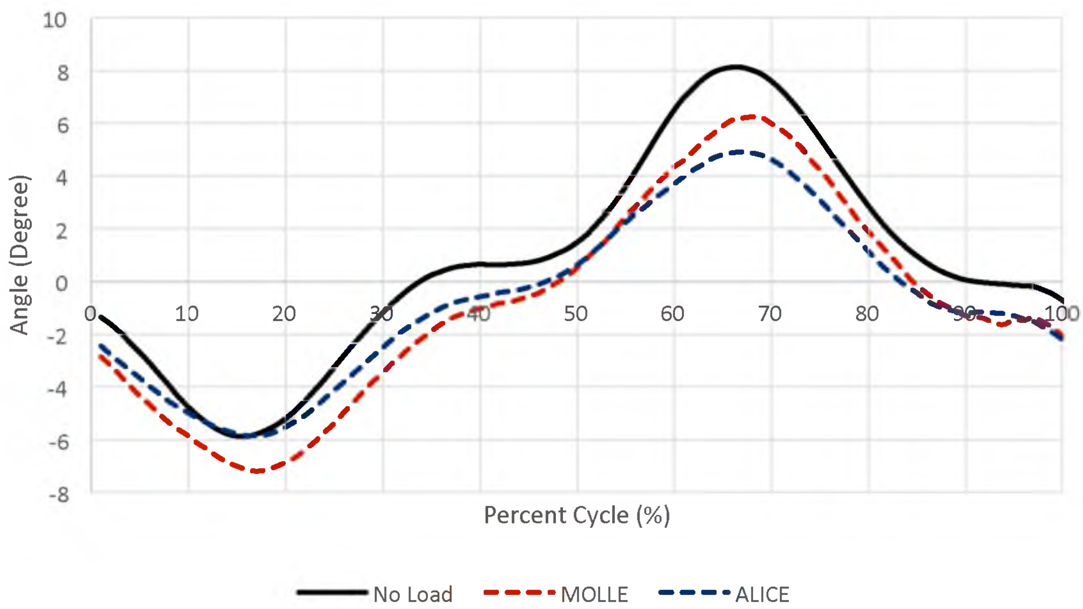

Figure 3-48. Thorax-Pelvis Lateral Bending Profile (Coronal)

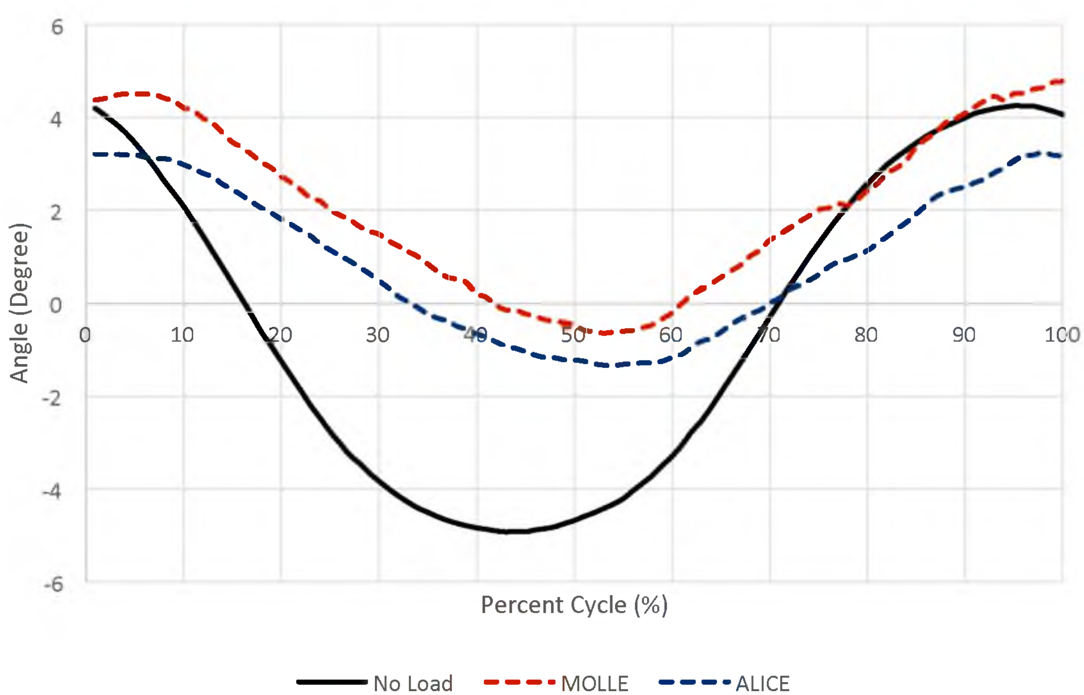

Figure 3-49. Thoracic Rotation Profile (Transverse)

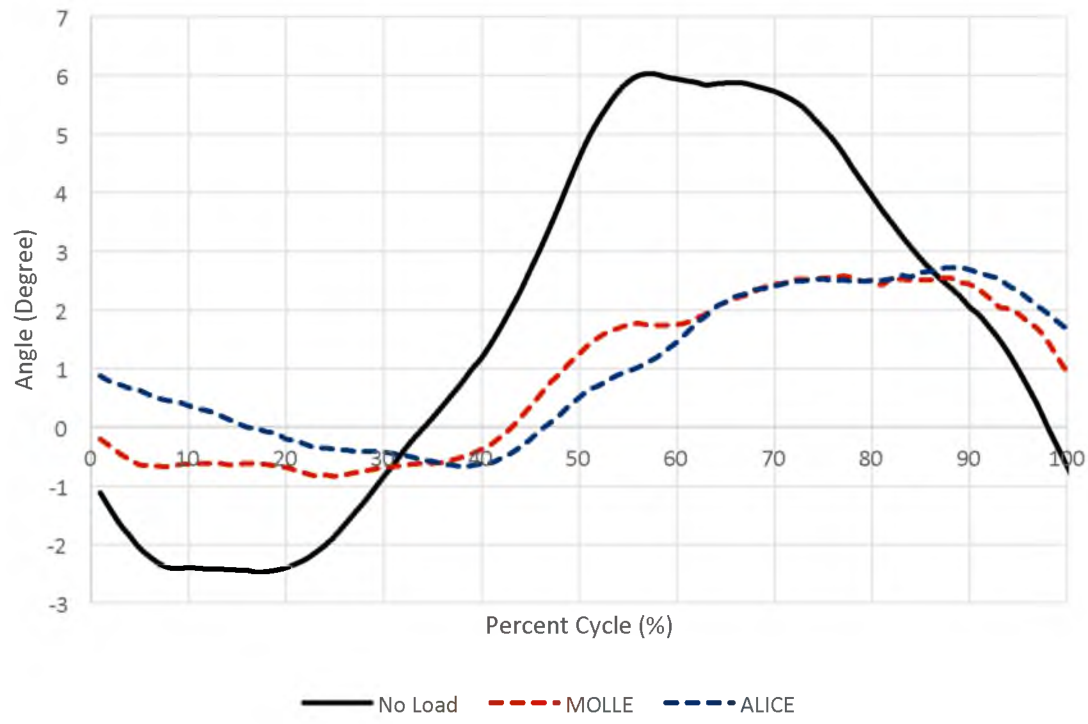

Figure 3-50. Pelvic Rotation Profile (Transverse)

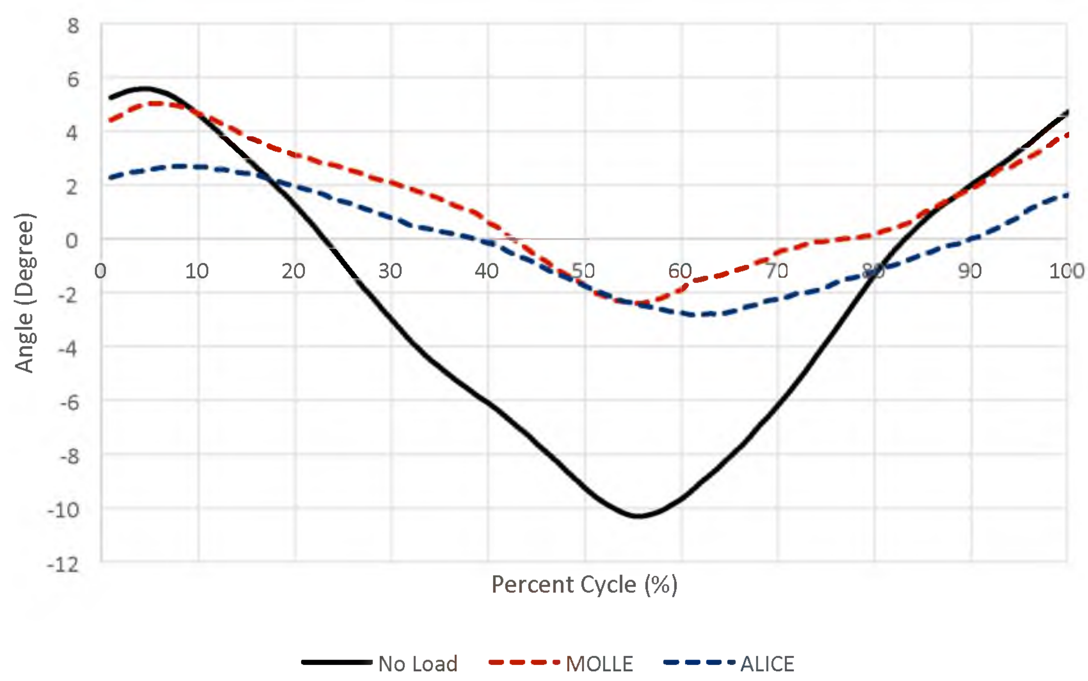

Figure 3-51. Thorax-Pelvis Rotation Profile (Transverse)

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CHAPTER 4

ERECTOR SPINAE MUSCLE ACTIVITY WHEN WALKING WITH A MILITARY BACKPACK

The purpose of this study was to identify the effects of surface and loading conditions on erector spinae muscle force activity from EMG data. There was a significant increase in muscle force on the sand surface than on the hard surface, and decrease in force with backpack carriage than with no load. Contralateral activation of the erector spinae muscle was found during each gait cycle. Peak muscle contraction occurred at each heel strike.

Introduction

Heavy load carriage is a risk factor for low back injury (Reynolds et al., 1990). Back pain in young people has been found to be related to heavily loaded backpacks (Korovessis et al., 2004; Negrini et al., 2002; Sheir-Neiss et al., 2003).

Although the current US army field manual recommends guidelines for the soldier's load on each operation (USFM21-18), soldiers typically carry loads exceeding the recommended guidelines (Knapik et al., 2004); the overall effect of these heavy loads on the soldier's body and the impact on the soldier's performance is still uncertain.

Reynolds examined injuries associated with maximum effort marching training.

While observing soldiers in marching training, 24 % of them had injuries such as a foot blisters, back pain, ankle sprains, and so on. Among these injuries, foot blisters (35 %) and back problems (23 %) were reported as the most common. These injuries may seem to be minor problems. In this study, however, 36 % of 218 soldiers suffered one or more injuries leading to 69 days of limited duty (Reynolds et al., 1999). In another study, 50 % of participants were unable to complete 20 km of strenuous marching due to problems associated with the back (Knapik et al., 2004). This is a huge noncombat loss to the commander.

The erector spinae is a large muscle of the back that originates near the sacrum and extends up the length of the back. It is essential to measure the muscle activity for calculating back compressive force (BCF) on the L5/S1 disc because its contraction or extension directly affects the magnitude of BCF.

Many researches use EMG for measuring erector spinae muscle activity (Bobet et al., 1984; Cholewicki et al., 2000; Dolan et al., 1993; Dolan et al., 1994; Dolan et al., 1995). In our study, surface EMG was used for detecting erector spinae muscle activation.

Method

EMG Setup and Preparation

Two channels of single differential surface EMG sensors (Delsys®) and the Bagnoli-8 Amplifier (Delsys®) were used. System gain was set as 1K for all data collection, and sampling frequency was 2000 Hz (Doerschuk et al., 1983).

A sensor was placed on the right erector spinae and the left, respectively. They were placed 20 cm above the PSIS markers and 2 cm lateral to the midline of the thorax (Cioni et al., 2010). The reference electrode was placed on the Processus mastoideus. The surface of the sensor was cleaned and skin preparation (shaving, cleaning) was performed before applying the sensor to the skin.

EMG Calibration

The target muscle must be isolated from other muscle activities in EMG calibration. The best way to isolate the erector spinae muscle is to confine hip movement while applying incremental load on the upper body. Thus, we proposed the custom calibration platform shown in Figure 4-1.

After confining pelvis movement using hip belts on the platform, 0 to 60 lbs of load was applied on the upper body of the subject using 10 lbs increments. EMG data were collected for 5 seconds while the subject was resisting the load. The subject's posture was captured during calibration data collection.

The 3DSSPP (3D static strength prediction program), also known as the University of Michigan Model, was adapted to estimate muscle force from a given static posture and load. From the applied load on the shoulders and the captured posture, the erector spinae muscle force was estimated from the model. Table 4-1 summarizes the calibration results.

EMG Data Processing

The linear envelope detection technique was used to extract information from the collected EMG waveform. All processes were based on the Kamen's data processing techniques (Kamen et al., 2010). A high-pass filter (30 Hz) was applied to raw EMG data for ECG (electrocardiographic) noise removal (Redfern et al., 1993). Full-wave rectification was then applied to the filtered data (Murray et al., 1985). Moving average, with 50ms of window size, was applied to obtain a smooth curve. A low-pass filter (10 Hz) was applied as a final step.

Data Analysis

The collected data were analyzed using SPSS (Ver.18.0, IBM Corporation, Armonk, NY) with a significance level of 0.05. Erector spinae muscle force was analyzed using MANOVA and post-hoc test. Gait cycle was normalized from 0 % (left heel strike) to 100 % (left heel strike). To estimate overall muscle forces we added left and right muscle forces under the normalized gait cycle.

Results

Erector Spinae Muscle Force

Figure 4-2 shows the average erector spinae muscle force. From MANOVA, erector spinae muscle forces showed statistical differences. A higher force was found on sand than on the hard surface ($p<.001$; Figure 4-3); at self-paced speed than at 4 km/h ($p=.01$; Figure 4-4); and walking with no load than when with a backpack load ($p<.001$; Figure 4-5). In addition, interactions between slope (flat, slope) and surface type (hard, sand, $p=.042$; Figure 4-6); slope and speed (4 km/h, self-paced, $p=.007$; Figure 4-7); and speed and backpack (no load, MOLLE, ALICE; $p=.010$; Figure 4-8) were significant.

Discussion

Effects of Backpack Carriage on Erector Spinae Muscle Force

Backpack load was a significant factor (*p*<.001). Table 4-2 summarizes the post-hoc analysis (Tukey LSD) results that identify the differences between the carriage conditions.

From the table, it can be seen that we found a significant difference between the no load and backpack loading conditions. No difference was found between MOLLE and ALICE.

We found that there was slight decrease in muscle force when carrying a backpack as we see in Figure 4-9. In our research, we only measured erector spinae muscle activations, however, we suspect that antagonistic co-contraction (van Dieën et al., 2005) of other upper body muscles for spinal stability may explain this decrease. It might also be related to change in upper body posture due to backpack carriage.

Motmans measured and compared trunk muscle activity in different modes of carriage such as with a backpack, a shoulder bag, a front pack, a double pack, and no bag. They found EMG levels of the erector spinae muscle were significantly lower when carrying a backpack, but also detected significant increase in rectus abdominis activation. They explained;

> "With no load, the back muscles must resist a trunk flexion moment because the centre of gravity of the upper body is located somewhat forward of the lumbosacral joint. With a load on the back, the combined centre of gravity of the trunk plus the pack shifts backward. This creates an extension moment. In order to counterbalance the weight on the back, a forward trunk lean occurs. A forward displacement can already be seen with loads less than 10 % BW. All these major shifts in body alignment can be interpreted as compensations to stabilize the whole-body centre of gravity over the feet. The net result of the rearward of the centre mass and the counterbalancing is a reduction in erector spinae activity" (Motmans et al., 2006).

Maruta also found decreased erector spinae muscle activation with forward torso flexion (Maruta et al., 2006). Additional research is needed in the future to identify the reason for decrease in muscle force.

Effects of Operational Terrains on Muscle Activation Pattern and Its Implication

METT-TC (mission, enemy, terrain, troops, time, and civilian) factors are key considerations for commanders when planning an operation. One of main goals of this study was to identify the effect of the terrain factor using quantitative method.

Figure 4-10 shows a muscle force profile on hard (solid line) and sandy surfaces (dotted line) when there is no load. Peak Force was recorded at 52 % of the gait cycle on both surfaces and there was about 50 % increase in peak muscle force on sand compared to the hard surface. Soldiers are often exposed to reconnaissance patrol during their mission with minimal equipment. This result implies there might be a higher possibility for back pain or muscle stress on desert terrain. Even though this result ruled out the effects of personal equipment (i.e., rifles, ammo, helmets and so on) during the mission, appropriate work-rest cycles need to be implemented based on operational terrain in order to increase their operational performance and injury prevention.

Figure 4-11 shows the back muscle force profile when carrying a backpack load. When manuevring with and without a backpack, sand terrain showed higher back muscle forces. This result can be applied to compare forces when soldiers march with full packs along desert (sand; dotted line) terrain versus urban (hard; solid line) terrain. There was 91 % increase in peak muscle force on the sand surface when carrying a backpack load compared to the hard surface. This would imply that the typical training (50 min.) – rest (10 min.) cycle in current guidelines should be modified for desert operations.

Figure 4-12 and Figure 4-13 illustrate right and left erector spinae muscle activation profiles. Generally, when there was a left heel strike, right muscle activation occurred and vice versa (contralateral activation). The left muscle activation started at about 40 % into the gait cycle, and maximized at around 50 %. A previous study also supports contralateral activation patterns of erector spinae muscles (Cioni et al., 2010). This activation pattern was clearer on the hard surface compared to the sandy surface because on the sand surface the other muscles (i.e., paravertebral muscles; left muscle in case of left heel strike) were also activated. This may be because erector spinae muscles activate more to maintain stability on the sand surface.

Lower limb kinematic analysis results showed increased ankle dorsi/plantarflexion, increased knee flex/extension, increased hip flex/extension and increased knee ab/adduction RoM angles when walking on the sand surface compared to the hard surface. These results also support the activation pattern on sand surface.

In Chevutschi and co-authors' research (Chevutschi et al., 2007), they compared erector spinae muscle activity on dry surface and in water. There were two clear bursts of erector spine muscle activity on dry ground, but more continuous activity with increased electrical activity was observed in water.

Their results in water show similarities to the sand surface results in our study. We assume that they are related because more vigorous stabilization processes are involved on irregular or uneven walking surfaces. However, additional research is needed to better explain the relationship.

Overall, two main risk factors were identified in this study for general walking on sand surface. One was increased amplitude of overall muscle force. The other was increased activation frequency of each muscle to maintain balance. The amplitude of the subactivation ratio on each heel strike (compared to main contraction of contralateral muscle) was 65 % with no load and 71.5 % with a backpack. This implies backpack carriage on a sandy surface can increase back muscle fatigue with prolonged exposure. However, additional research is required to identify the muscle fatigue from increased activation frequency.

Upper Body Stabilization Mechanism: The Relationship between Thoracic Flexion and Pelvic Tilt and Erector Spinae Muscle Activity

In this study, we found a significant increase in thoracic flexion and pelvic anterior tilt with backpack carriage. It is well known that erector spinae muscles play a significant role in thoracic and pelvic movement (Kang et al., 2013). Pelvic tilt occurs as a result of erector spinae muscle contraction as is illustrated in Figure 4-14.

Framed packs exert a consistent anterior force on the lower back, and thus it has been suggested that this force could contribute to low back pain and soreness (Lafiandra et al., 2004).

Figure 4-15 shows the results from one of the trials (subject# 20, MOLLE, hard surface, 4 km/h, 3rd trial). Figure 4-15A illustrates the thoracic flexion profile; positive degrees represent forward sagittal flexion. Figure 4-15C illustrates pelvic tilt; positive degrees mean pelvic anterior tilt in the sagittal plane. Figure 4-15B shows the on-off timing of the right (lighter color) and the left (dark color) erector spinae muscles. DelSys EMGworks Analysis (Version 4.1.1.0) was used for calculating erector spinae muscle on-off timing and the X-axis represents one gait cycle in terms of time (second). In this trial, one gait cycle was 1.16 seconds. Peak thoracic flexion occurred at the time of right heel strike (i.e., 0.59 seconds), and then muscle contraction was detected at 0.663 seconds. Peak pelvic anterior tilt followed at 0.67 seconds. This result supports the consequential stabilization mechanism of the thorax, erector spinae muscle, and pelvis. Further investigation, however, with a larger sample size is required for generalization.

Pope et al. hypothesized that the muscle would show greater reaction time latency and a larger response amplitude when faced with a sudden load after whole body vibration exposure (1998). Also, Rohlman et al. (2006) explained that "muscle forces stabilize the spine and have a great influence on spinal load" and that upper body flexion is most likely combined with bending of the spine. In this regard, the increased muscle reaction time latency would induce delayed pelvic adjustment. It might thus increase the duration of uneven concentration of pressure on L5/S1 disc. Consequently, it would result in a higher potential for low back pain or disc failure. Further research is needed to quantify the effect of the muscle reaction time latency after prolonged exposure to whole body vibration.

Lamoth et al. (2006) hypothesized that alteration of trunk-pelvic coordination and erector spinae muscle activity timing occur to minimize the effect of unexpected perturbation. In their research, they found that the trunk-pelvic coordination was velocity (walking speed) dependant. Our research only had two speed conditions, but we might be able to find the effect of walking speed with extended analysis in the future.

Conclusion

When walking with and without a backpack, sand terrain created higher back muscle forces and a contralateral activation pattern of erector spinae muscles was more distinct on the hard surface compared to the sandy surface. This is because the other back muscles (i.e., paravertebral muscles; left muscle in case of left heel strike) were also activated on the sand surface than on the hard surface. This may be because erector spinae muscles activate more to maintain stability on the sand surface.

Overall, two main risk factors were identified in this study for general walking on a sand surface. One was increased amplitude of overall back muscle force. The other was increased activation frequency of each erector spinae muscle to maintain balance.

There was a slight decrease in erector spinae muscle force when carrying a backpack. We suspect that antagonistic co-contraction of other upper body muscles for spinal stability may explain this decrease. We only measured erector spinae muscle activations in our research. Thus, additional research is required to identify the co-contraction of upper body muscles in the future.

Figure 4-1. Custom EMG Calibration Platform

Table 4-1. EMG Calibration Result

	Left Erector Spinae			Right Erector Spinae		
Subject #	Sensitivity (lbs/volt)	Offset (volt)	Linearity (R-squrare)	Sensitivity (lbs/volt)	Offset (volt)	Linearity (R-squrare)
S001	1307.7	35.202	0.981	1304.8	24.502	0.971
S002	1890.5	19.818	0.913	1702.4	25.999	0.95
S003	1907.6	54.176	0.953	1469.7	16.747	0.851
S004	980.16	43.753	0.885	1718.7	41.826	0.831
S005	1677	41.24	0.994	1867.9	46.284	0.978
S006	598	54.222	0.934	543.6	44.923	0.961
S007	1024.6	42.507	0.971	718.19	52.56	0.932
S008	1322.3	31.488	0.979	1501.3	26.309	0.88
S009	1500.2	37.72	0.991	1809	45.996	0.98
S010	731.05	45.444	0.977	1124.4	41.17	0.984
S011	1073.8	58.947	0.932	1016.8	44.289	0.985
S012	333.46	37.276	0.972	244.55	40.949	0.992
S013	739.93	49.474	0.944	646.68	37.491	0.993
S014	1502.9	45.313	0.99	1288.7	41.818	0.989
S015	1875.1	44.566	0.992	1799.4	42.375	0.981
S016	772.28	50.236	0.98	877.4	52.539	0.972
S017	674.56	48.249	0.988	756.57	53.372	0.958
S018	1172.4	64.201	0.844	1865.8	58.504	0.901
S019	1247.9	62.514	0.942	1463.8	53.468	0.978
S020	2208.2	51.318	0.928	1844	48.338	0.927

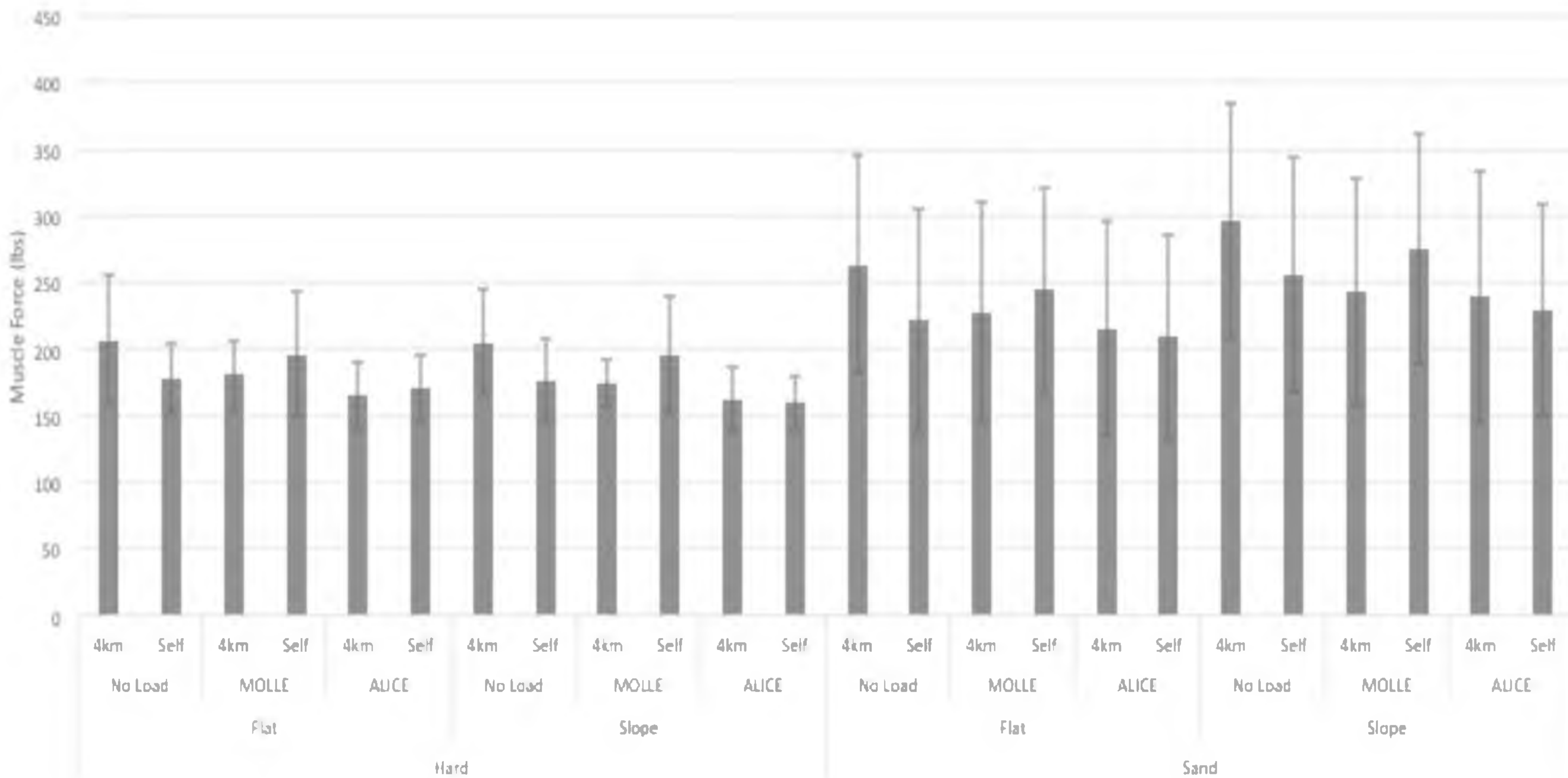

Figure 4-2. Erector Spinae Muscle Force

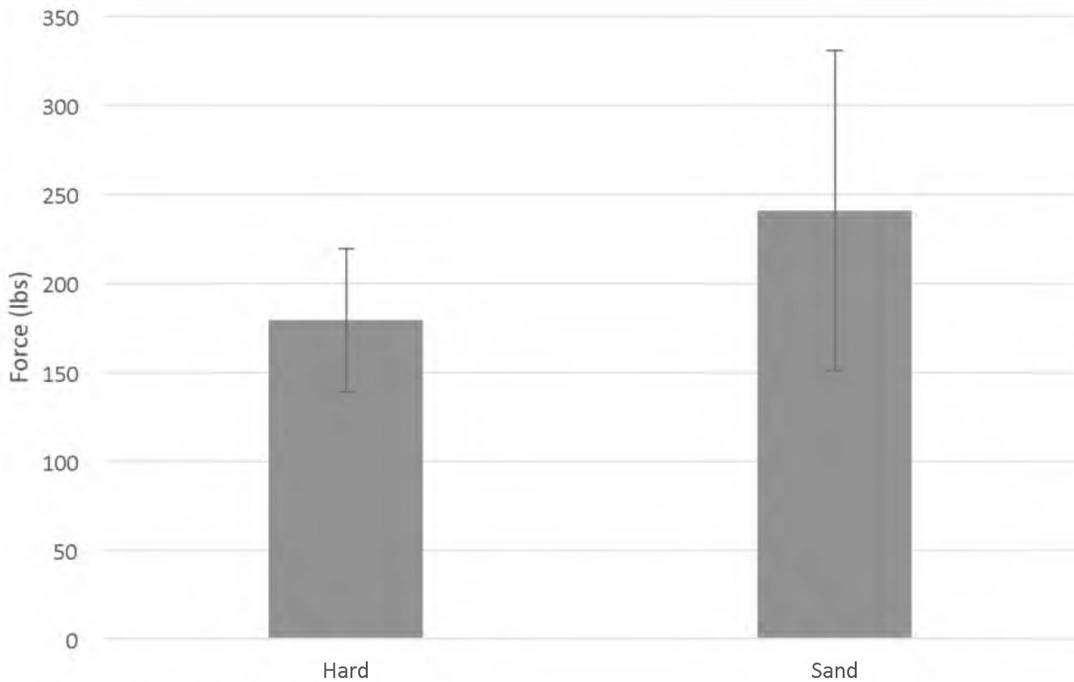

Figure 4-3. Erector Spinae Muscle Force by Surface ($p<.001$)

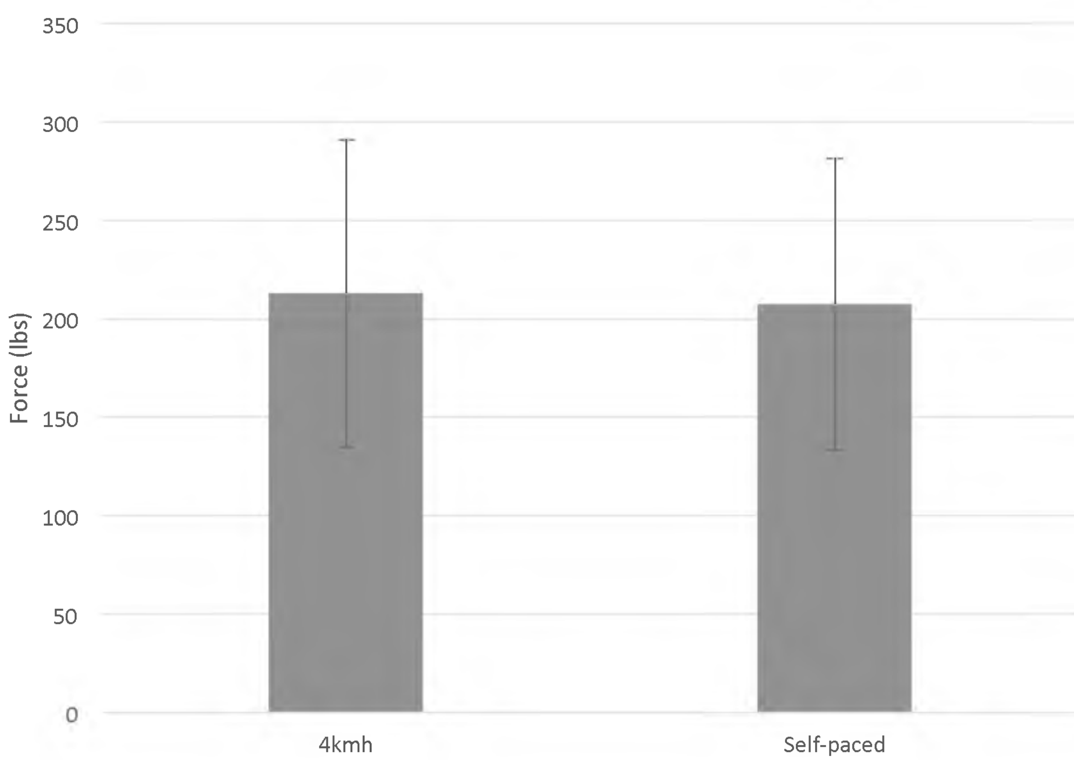

Figure 4-4. Erector Spinae Muscle Force by Speed ($p=.01$)

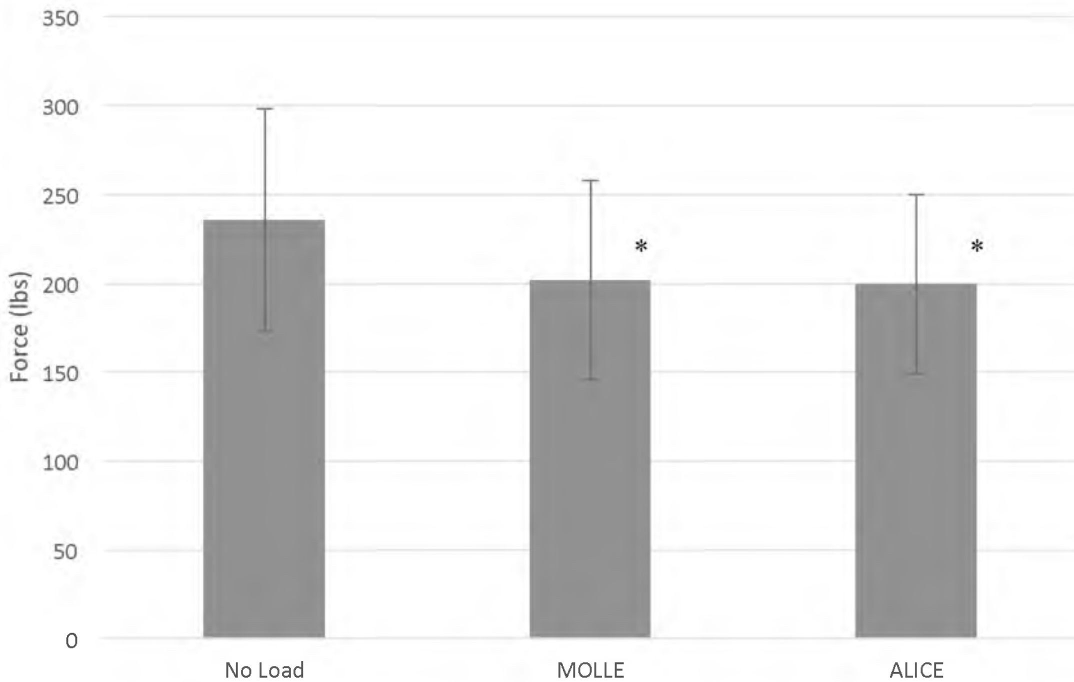

Figure 4-5. Erector Spinae Muscle Force by Backpack ($p<.001$)
* indicates statistical significance ($p<.005$) compared to No Load

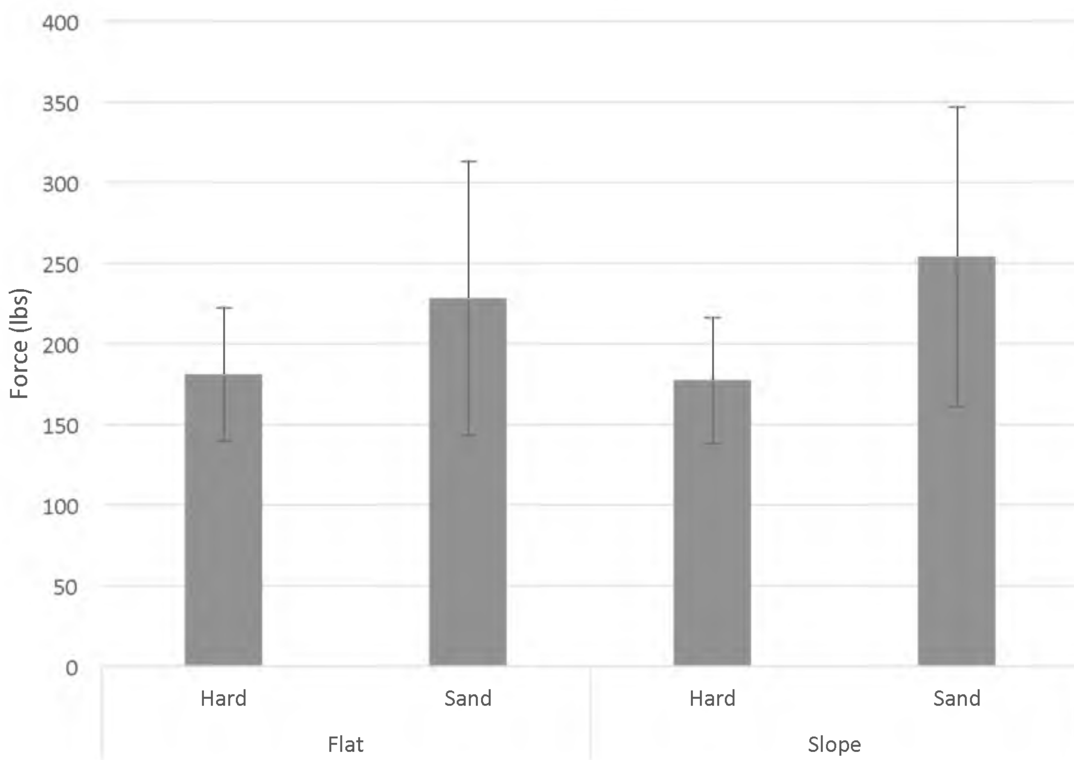

Figure 4-6. Erector Spinae Muscle Force by Slope and Surface ($p=.042$)

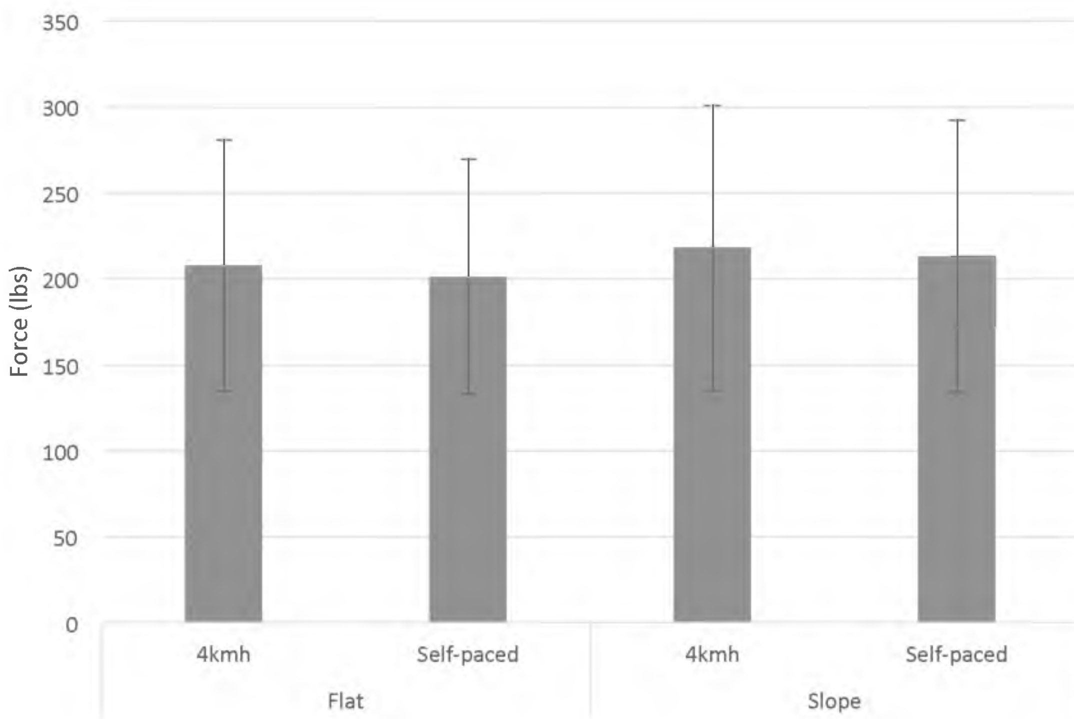

Figure 4-7. Erector Spinae Muscle Force by Slope and Speed (p=.007)

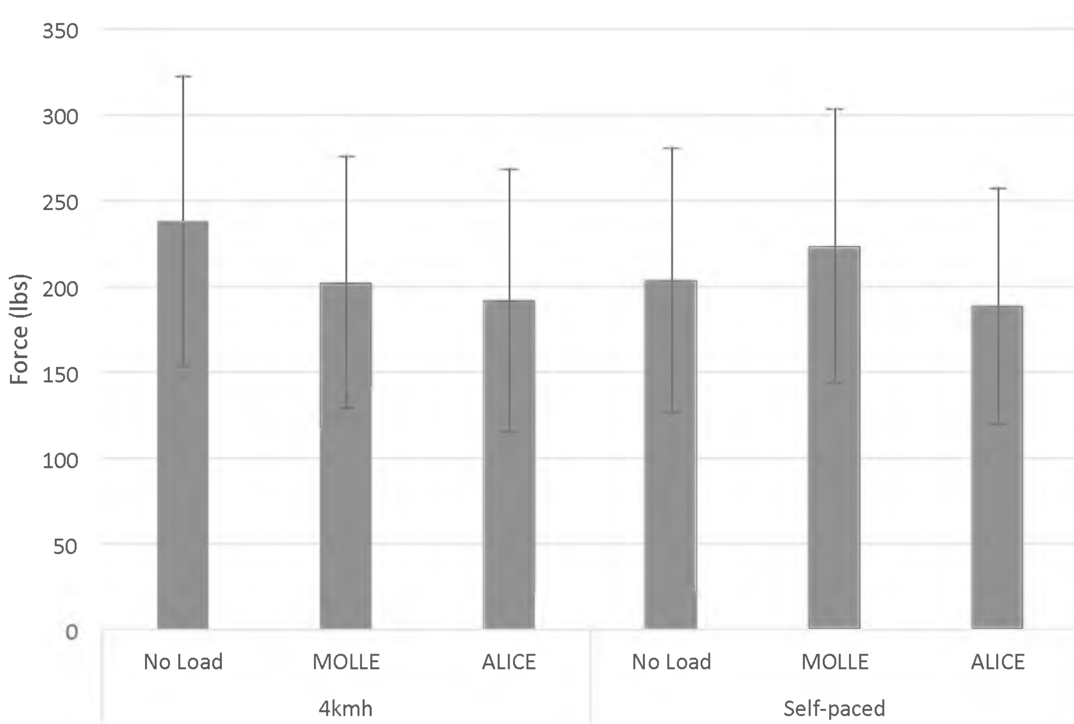

Figure 4-8. Erector Spinae Muscle Force by Speed and Backpack (p=.01)

Table 4-2. Effect of Backpack Carriage on Erector Spinae Muscle Force

Post hoc		
No Load vs. MOLLE	No Load vs. ALICE	MOLLE vs. ALICE
0.000	0.000	0.327

Note: Numbers in the table represent *p* values.

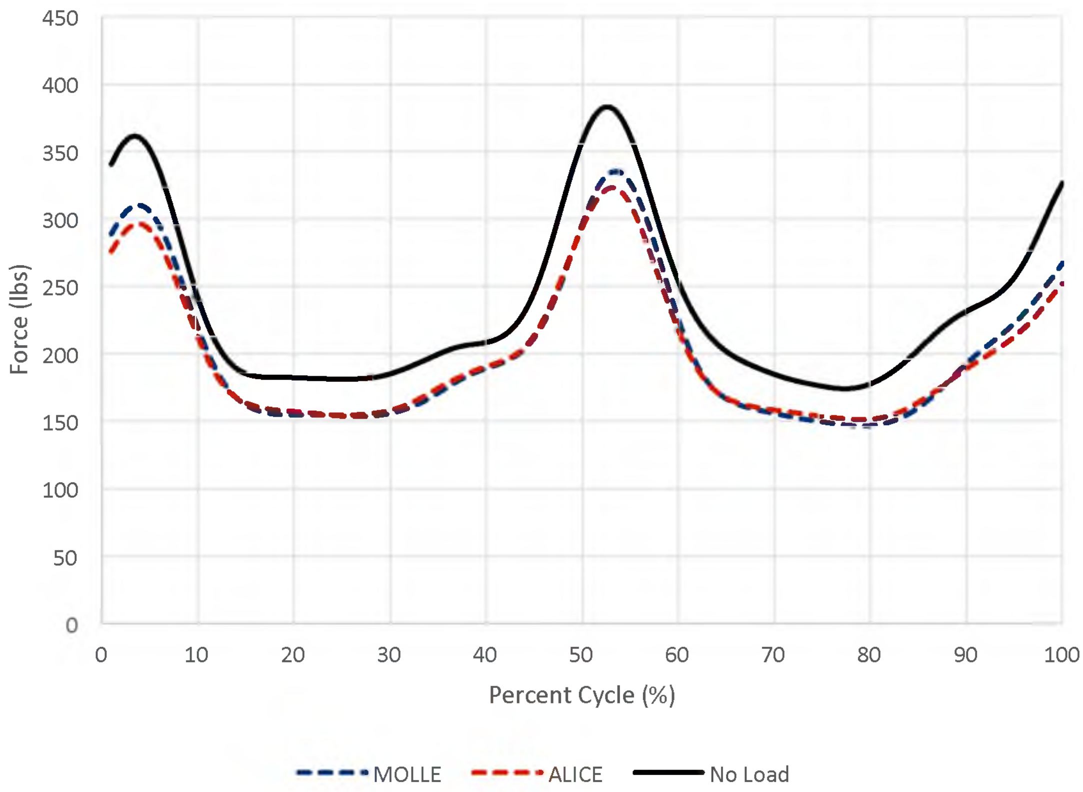

Figure 4-9. Erector Spinae Muscle Force Profile

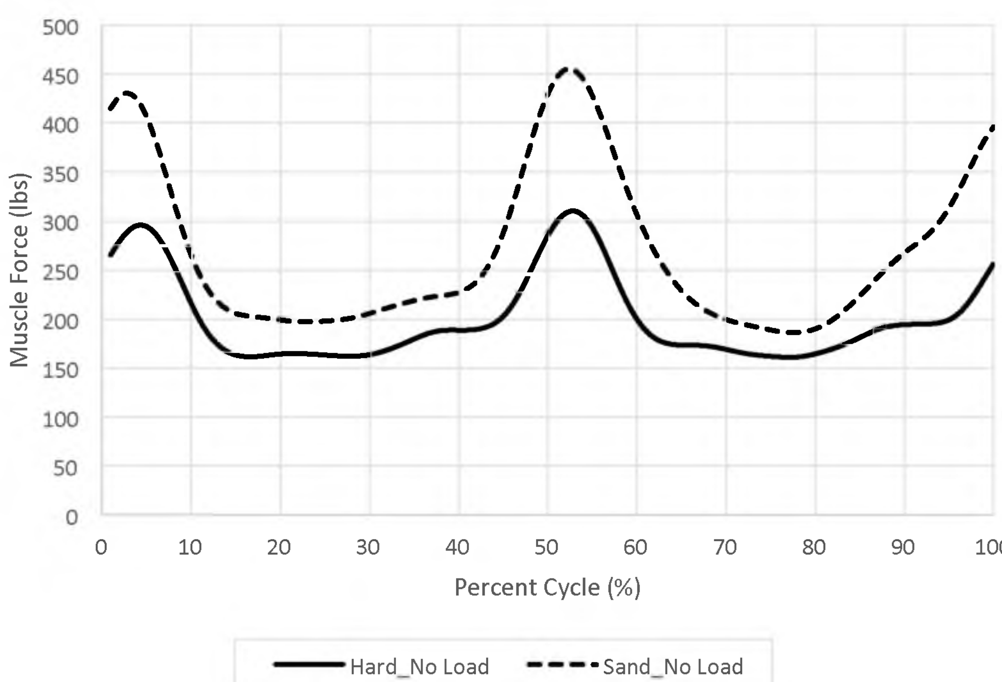

Figure 4-10. Erector Spinae Muscle Force Profile (no load)

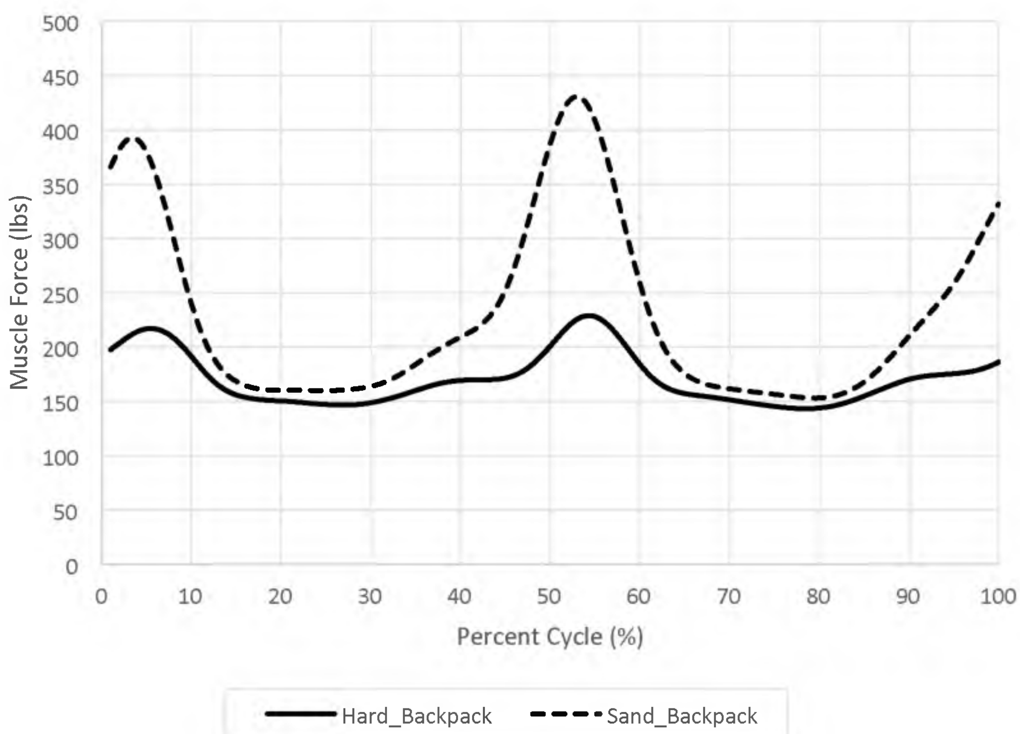

Figure 4-11. Erector Spinae Muscle Force Profile (with backpack)

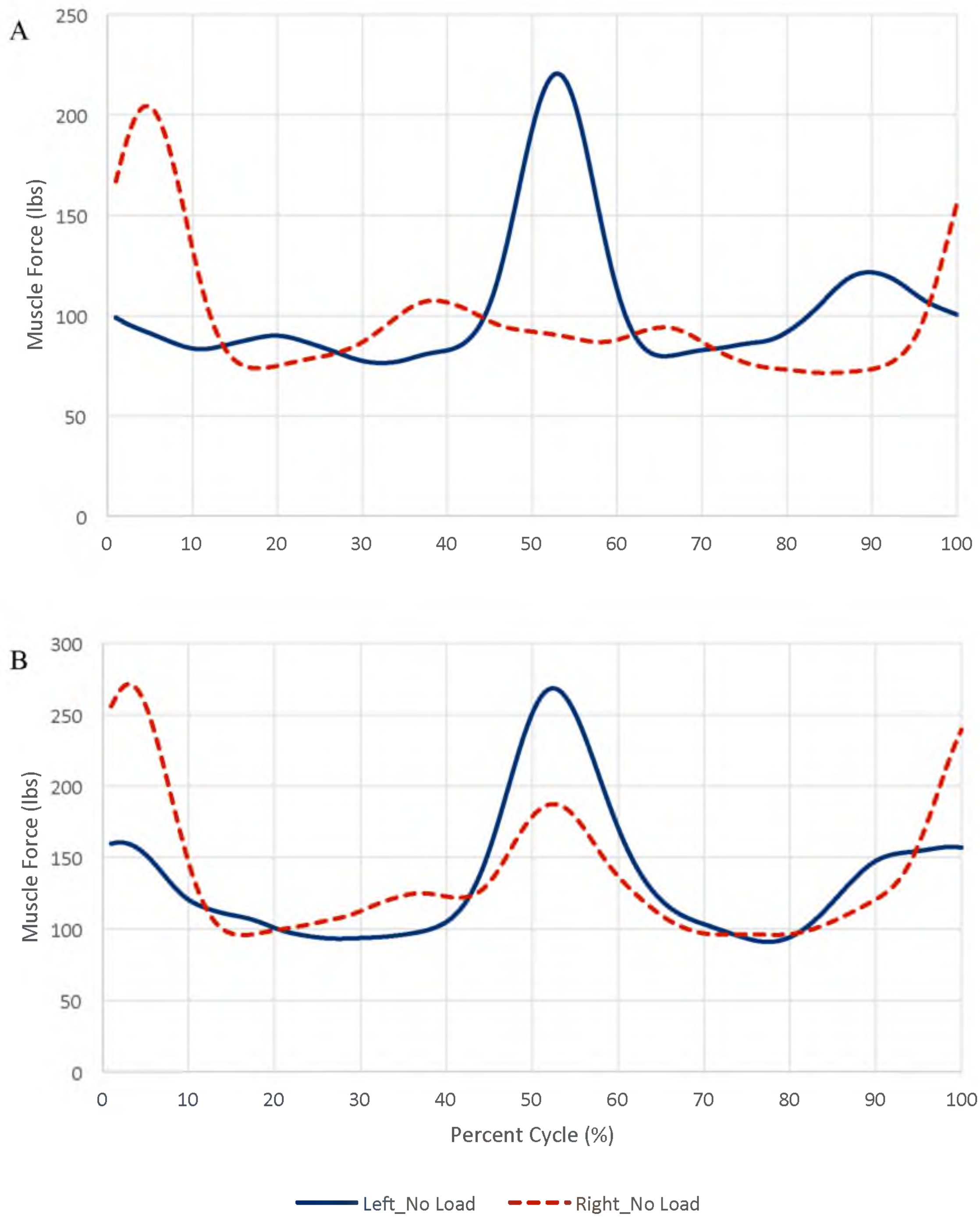

Figure 4-12. Back Muscle Activation Profile (no load)
A) Hard Surface; B) Sand Surface

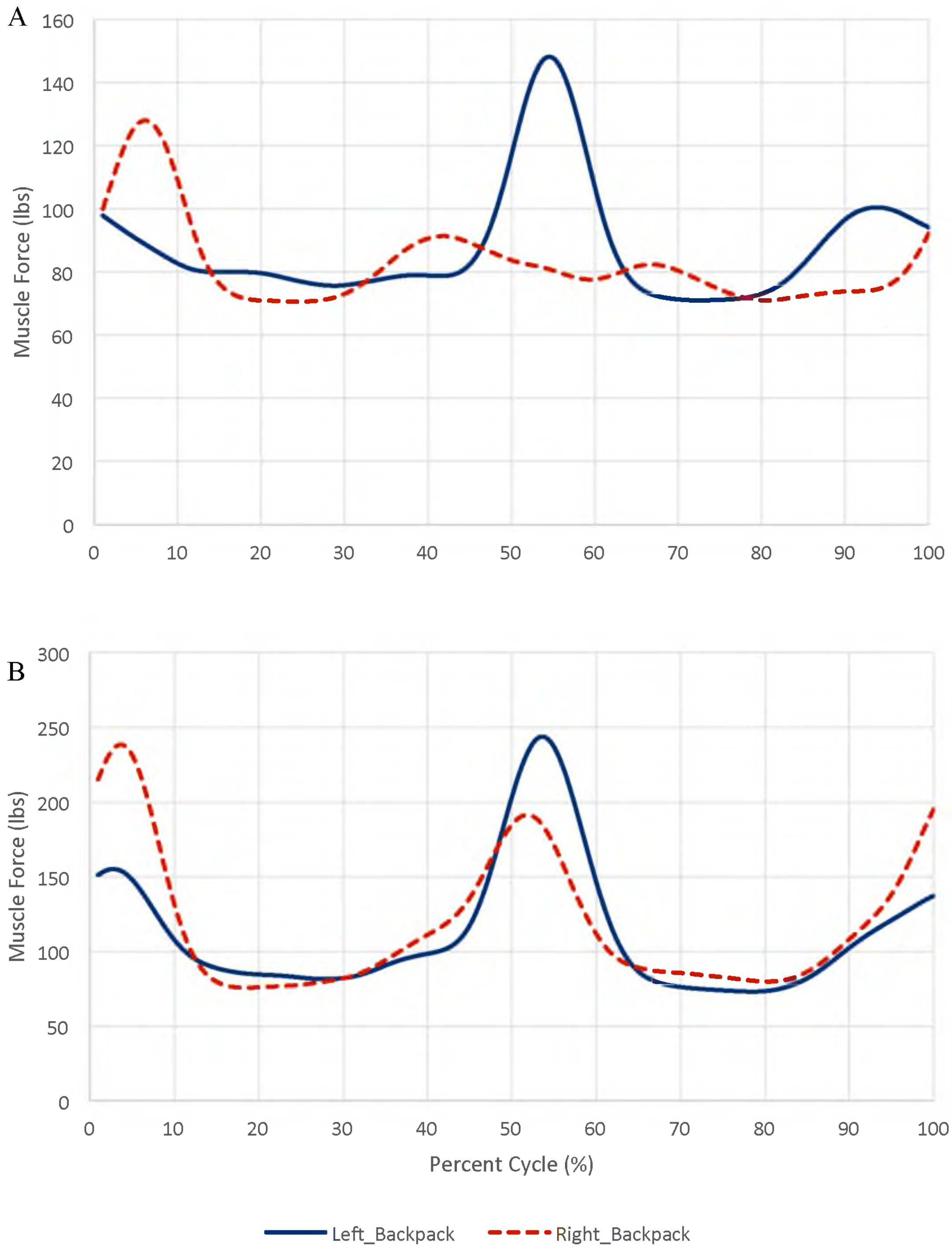

Figure 4-13. Back Muscle Activation Profile (with Backpack)
A) Hard Surface; B) Sand Surface

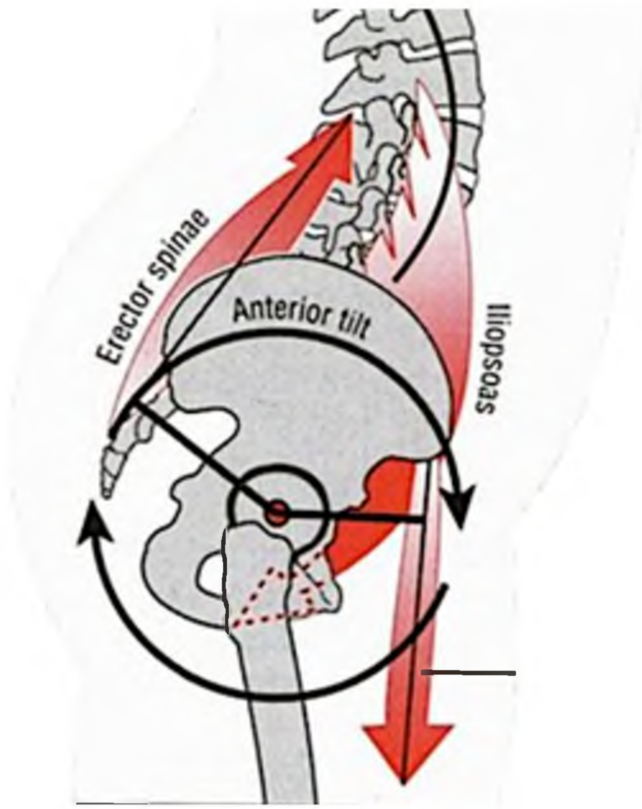

Figure 4-14. Relationship between Muscle Contraction and Pelvic Tilt

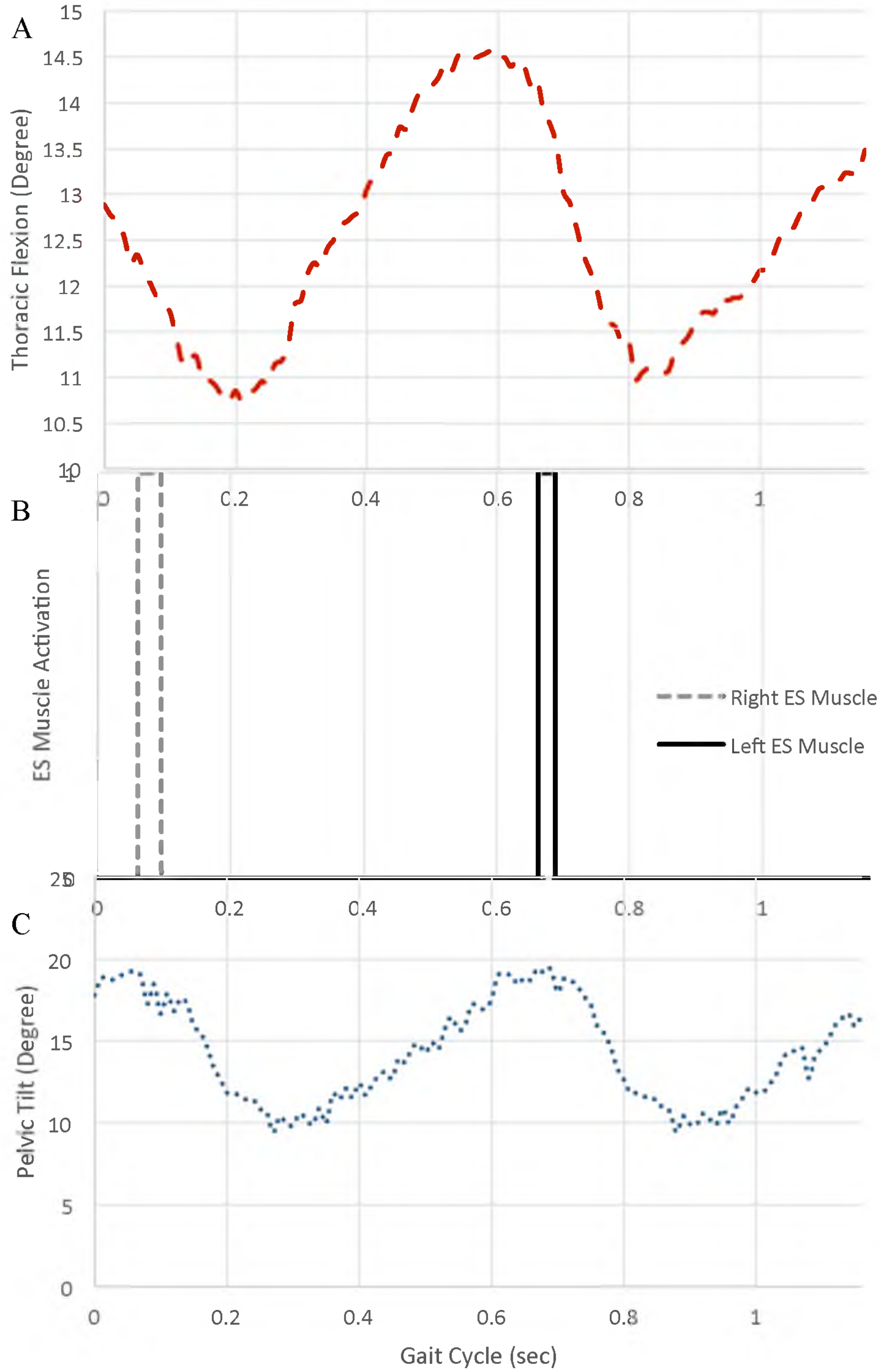

Figure 4-15. Upper Body Stabilization Mechanism
A) Thoracic Flexion Angle; B) Muscle Activation Time; C) Pelvic Tilt Angle

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CHAPTER 5

BIOMECHANICAL ANALYSIS OF BACKPACK DESIGN: FOCUSING ON FORCES ON SHOULDERS AND LOW BACK AND THEIR IMPLICATIONS

The purpose of this chapter is to use biomechanical models to analyze the shoulder reaction forces and the low back contact force that result from backpack load. In this study, customized load cells were fabricated to measure the shoulder force. They revealed a higher shoulder reaction force on the right shoulder than on the left shoulder; this might be related to the thorax coronal motion. Comparing backpack types, the MOLLE design resulted in a higher shoulder reaction force than the ALICE design. The ALICE, in turn, produced a higher contact force on the low back than the MOLLE. Overall, proper load distribution must be considered when designing a backpack in order to meet pain perception criteria.

Introduction

Shoulder pain is associated with wearing heavy backpacks (Macias et al., 2008). Thus, it is proposed that shoulder pressure is a significant issue in backpack design. Shoulder straps of backpacks exert pressure on the skin and, in general, this pressure is higher with a frameless pack than with a pack with a frame and hip belt (Knapik et al., 2004). Al-Hazzaa investigated the relationship between pain experienced by students using backpacks and their manner of backpack carriage. It was discovered that more than one-third of students have experienced this pain. Shoulder pain was the highest pain reported by the students (33.2 %), being more predominant even more so than back pain when carrying backpacks (2006).

According to the research of Vaheron et al., shoulder pressure is a limiting factor in heavy backpack carriage. Using four sensors per each shoulder (e.g. two anterior and two posterior) they measured the shoulder pressure fluctuation when subjects carried a backpack. They found that expert hikers experienced lower forces along the straps than did a novice group by assuming specific body postures and bending their upper body. They suggest the optimal way to reduce shoulder strain is for the backpack user to reduce stride length, wear appropriate footwear, and maintain their center of mass over their feet (1999).

A Canadian research group insists that maintaining low shoulder pressure while wearing a backpack is important since 90 % of soldiers report discomfort at 20 kPa of pressure. Specific pain perception levels based on shoulder reaction forces have been suggested to help with such maintenance (Reid et al., 1997).

Method

Load Cell Design and Fabrication

The physical dimensions of the load cell were optimized using Pro Engineer Software (Figure 5-1). Aluminum 6061 was chosen for the base material. The prototypes were cut using a Water Jet cutting machine (OMAX 2626 JetMachining Center) and manual milling (Kurt Mfg. Co. Model D675) processes (Figure 5-2).

Two sets of strain gauge (SGT-1/350-XY13) per each sensor were used to build the Wheatstone bridge circuit. INA122 was used as an amplifier and the circuit was designed based on the datasheet provided (Figure 5-3).

Load Cell Calibration

This load cell is designed to measure the axial force in the shoulder strap. To determine the load cell characteristics, a custom calibration test fixture was created (Figure 5-4).

Five calibration trials were performed at room temperature for each load cell. Applied loads ranged from 0 to 75 as. with 25 lbs. increments. Sensitivity and linearity were derived from the collected data and the load-voltage relationship was established using least squares regression techniques. Mean offset was 1.092 voltages, and an offset drift (± 0.03 v) was identified. This drift could have caused unwanted error on each measurement, so we used mean-subtraction from the raw data in every trial to remove the drift effect. Table 5-1 shows the calibration result.

Two markers were attached on each load cell to help determine the directional components of the force vector (Figure 5-5).

From this design and calibration process, the load cell was used to calculate shoulder reaction forces while carrying a backpack. From this collected data, shoulder reaction force fluctuations were identified and back compressive forces were estimated.

Biomechanical Model

Canadian forces (Pelot et al., 1998) have developed models to determine the reaction forces on the hip and shoulders while carrying a backpack (Figure 5-6).

In the study of Pelot et al., they estimated shoulder reaction force and low back contact force to compare whether these are within pain perception levels (1998). The first step to build the models to determine this is to simplify the upper body and backpack. While carrying a backpack, two main reaction forces are acting on users' upper body: the force on the shoulders (S^N) and low back contact force (F_x; as we see in Figure 5-6).

From the free body diagrams, F_x can be calculated from the following equations.

$$F_x = W \sin\beta + T_{1R}\cos\theta_{1R} + T_{2R}\cos\theta_{2R} + T_{1L}\cos\theta_{1L} + T_{2L}\cos\theta_{2L}$$

By measuring the tension in the upper shoulder strap (Right: T_{1R}, Left: T_{1L}) and lower shoulder strap (Right: T_{2R}, Left: T_{2L}) using the customized load cells, shoulder reaction forces (S^N) can be calculated from the following equations.

$$S^N = \sqrt{(S_x^N)^2 + (S_z^N)^2}$$

$$S_x^N = T_{1R}\cos\theta_{1R} + T_{2R}\cos\theta_{2R} + T_{1L}\cos\theta_{1L} + T_{2L}\cos\theta_{2L}$$

$$S_z^N = T_{1R}\sin\theta_{1R} + T_{2R}\sin\theta_{2R} + T_{1L}\sin\theta_{1L} + T_{2L}\sin\theta_{2L}$$

Data Processing and Analysis

Mean subtraction was performed from the raw analog data to remove offset, and then a low-pass filter (10 Hz) was applied to all digital data (from marker movement) and analog data (from voltage reading) that was compiled in Visual3D software. Afterward, a calibration curve was then applied using MATLAB (Version 2013b) in order to convert voltage to pound.

The collected data were analyzed using SPSS (Ver.18.0, IBM Corporation, Armonk, NY) with a significance level of 0.05. Shoulder reaction force and low back contact force were analyzed using MANOVA. The gait cycle was normalized from 0 % (left heel strike) to 100 % (left heel strike).

Results

Shoulder Reaction Force

Figure 5-7 summarizes averaged shoulder reaction forces (S^N). Shoulder reaction forces involved with the different variables showed statistical differences. Higher shoulder forces were found with MOLLE than with ALICE (p<.001; Figure 5-8); on sand than on the hard surface (p<.001; Figure 5-9); and at 4 km/h than at self-paced speed (p=.003; Figure 5-10). In addition, interactions between slope (flat, sloped) and backpack (no load, MOLLE, ALICE; p=.019; Figure 5-11); speed (4 km/h, self-paced) and backpack (p=.048; Figure 5-12); surface type (hard, sand) and slope (p=.005; Figure 5-13); and slope and speed (p=.012; Figure 5-14) were also significant.

Low Back Contact Force

Figure 5-15 summarizes averaged low back contact forces (F_x), which showed statistical differences. Higher forces were found with ALICE than with MOLLE (p<.001; Figure 5-16); on sand than on the hard surface (p<.001; Figure 5-17); and on the sloped surface than on the flat surface (*p*=.005; Figure 5-18). In addition, the interaction between surface type and speed (*p*=.009; Figure 5-19) was also significant.

Discussion

Shoulder Reaction Force and Low Back Contact Force Profiles

Profiles of shoulder reaction force (Figure 5-20) and low back contact force (Figure 5-21) were similar for both backpacks. However, in terms of force magnitude, the MOLLE resulted in significantly higher forces on the shoulder and lower forces along the low back when compared to the ALICE.

Peak forces were identified around 10 % and 60 % into the gait cycles. Figure 5-22 shows relative movement of center of mass (COM) of the backpacks relative to thoracic motion in the sagittal plane. When there was a significant downward movement of the COM, the shoulder reaction force and the low back contact force were increased (i.e., 0 % to 10 % and 50 % to 60 % of the gait cycles). The overall profiles may be related to relative movement of a backpack within a gait cycle. Additional analysis is necessary in the future to generalize the relationship between the COM displacement and the force fluctuations.

Pain Perception Level and Evaluation of Backpack Design

Pelot et al. proposed that the acceptable backpack load should satisfy two different pain perception levels to minimize pain on shoulders and the low back: 289 N for the shoulder reaction force and 135 N for the low back contact force (1998).

Both backpacks satisfied the shoulder criteria, but did not satisfy the low back criteria, as seen in Table 5-2.

Based on our results, a given load of 68 lbs is too heavy to fall within the pain perception level. Possible suggestions for load carriage might thus include reducing of the total load and modifying the load distribution between the shoulder and hip area.

There were some limitations in our study. Components of MOLLE (i.e., the assault pack and side pouches) were removed and the hip belt was modified to prevent ASIS marker occlusion. Additional research with different backpack weights and load distribution is needed to analyze the effects of load distribution in the future.

Relationship Between Shoulder Reaction Force and Thoracic Motion

The asymmetry profile of thoracic motion was more evident when carrying backpacks as we discussed in Chapter 3. This profile might be one of the significant factors affecting shoulder reaction force and discomfort.

Figure 5-23 illustrates the relationship between the shoulder reaction force profile and thoracic lateral bending in the coronal plane (S001, HF1S: Subject #001, Hard / Flat surface, MOLLE, Self-paced speed, 1^{st} trial). Figure 5-23A shows the left shoulder reaction force profile (i.e., the solid line) and the right shoulder's profile (i.e., the dotted line). Figure 5-23C represents thoracic motion in the coronal plane. In this graph, the right shoulder is higher than the left shoulder when the angle is negative, as is seen in the schematic drawing of the posture in the figure. The subject had a tendency to lift his right shoulder when carrying a backpack during the gait cycle and a neutral shoulder posture was identified at 60 % of the gait cycle. Previous research also showed the asymmetric shoulder motion in the coronal plane but they did not point out this tendency in their results (Bartonek et al., 2002; Linley et al., 2010; Nguyen et al., 2004). We found that the left and right shoulder reaction forces were identical at 60 % of the cycle. From 60 % to 80 % of the cycle, then the left shoulder force dropped as his left shoulder went down in the gait cycle region. As the result of the thoracic coronal bending, a higher shoulder reaction force was recorded as seen in Figure 5-24. The right shoulder force was significantly higher for both backpacks. Macias's research team also reported a higher shoulder pressure on the right shoulder and addressed the potential relationship between the shoulder pressure and the posture (2005, 2008). The difference between forces on each shoulder per backpack was 16.2 % (MOLLE) and 12.9 % (ALICE), respectively. MOLLE thus showed a greater imbalance of shoulder force than ALICE; this might be explained from a larger variation of thoracic motion when carrying the MOLLE pack (Figure 5-25), however, further study is required to clarify the relationship between the thorax motion and the shoulder reaction force in the future.

Conclusion

Backpack load exerted stresses on the shoulders. Profiles of shoulder reaction force and low back contact force were similar for both backpacks. However, in terms of force magnitude, the MOLLE resulted in significantly higher forces on the shoulder and lower forces along the low back when compared to the ALICE.

Based on our results, a given load of 68 lbs is too heavy to fall within the pain perception level. Possible suggestions for load carriage might thus include reducing the total load and modifying the load distribution between the shoulder and hip area. However, components of MOLLE (i.e., the assault pack and side pouches) were removed and the hip belt was modified to prevent ASIS marker occlusion in our study. Thus, additional research with all MOLLE pack components is needed to generalize and analyze the practical effects of load distribution in the future.

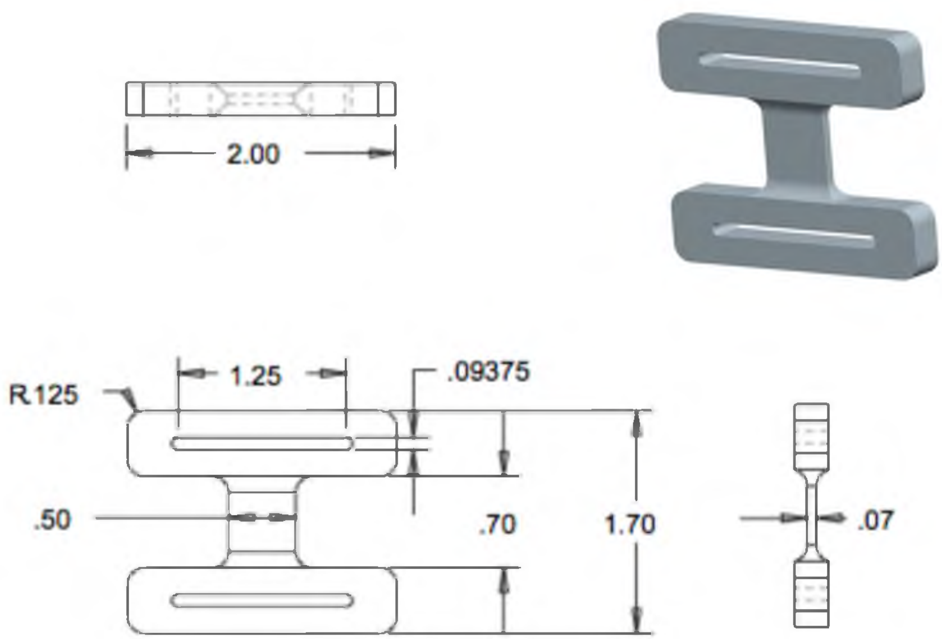

Figure 5-1. Load Cell Modeling

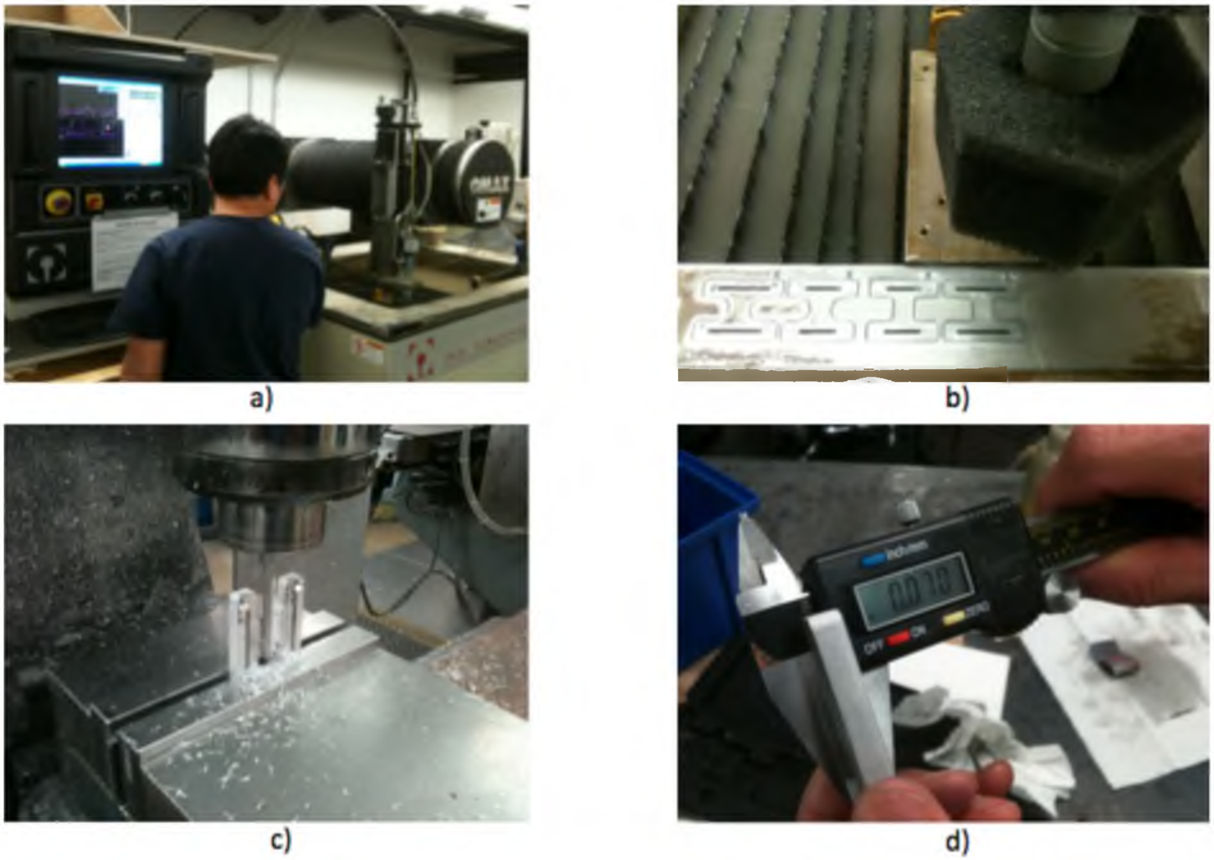

Figure 5-2. Machining Processes.
A) Set up; B) Water Jet; C) Milling; D) Deburring & Check

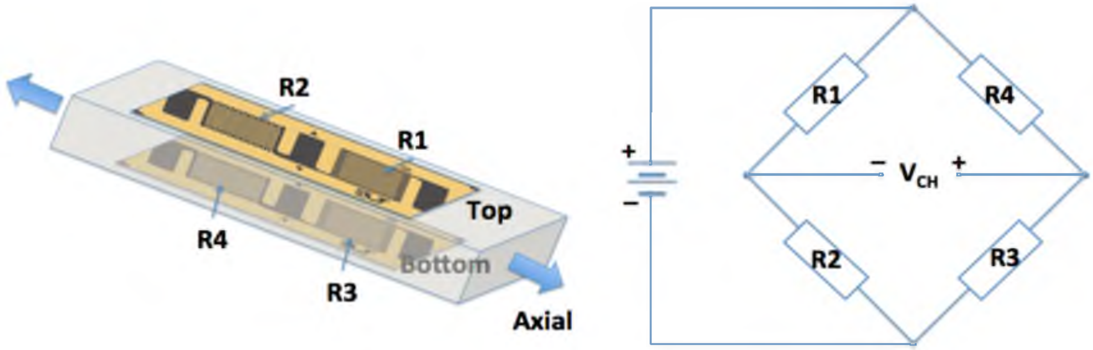

Figure 5-3. Wheatstone Bridge Configuration

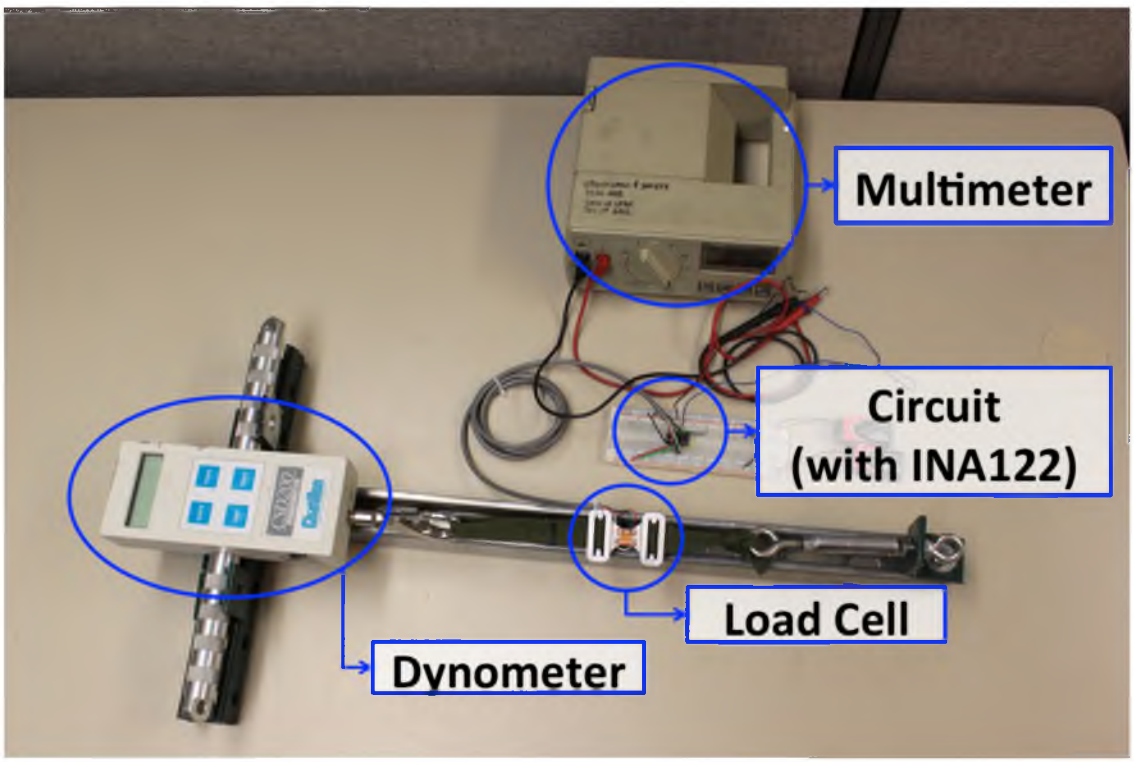

Figure 5-4. Calibration Test Fixture and Environment

Table 5-1. Load Cell Calibration Results

Sensor #	Sensitivity (lbs/volt)	Linearity (R-squrare)
LC1	23.256	0.996
LC2	22.439	0.994
LC3	22.616	0.999
LC4	17.999	0.977
LC5	26.746	0.999
LC6	27.105	0.999
LC7	20.728	0.999
LC8	30.096	0.989

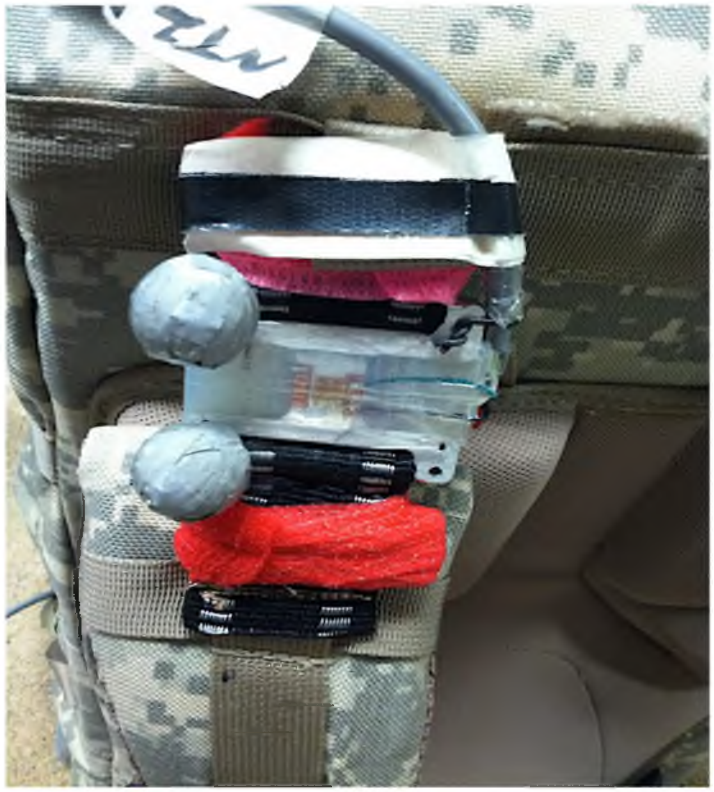

Figure 5-5. Load Cell Installation (with Markers)

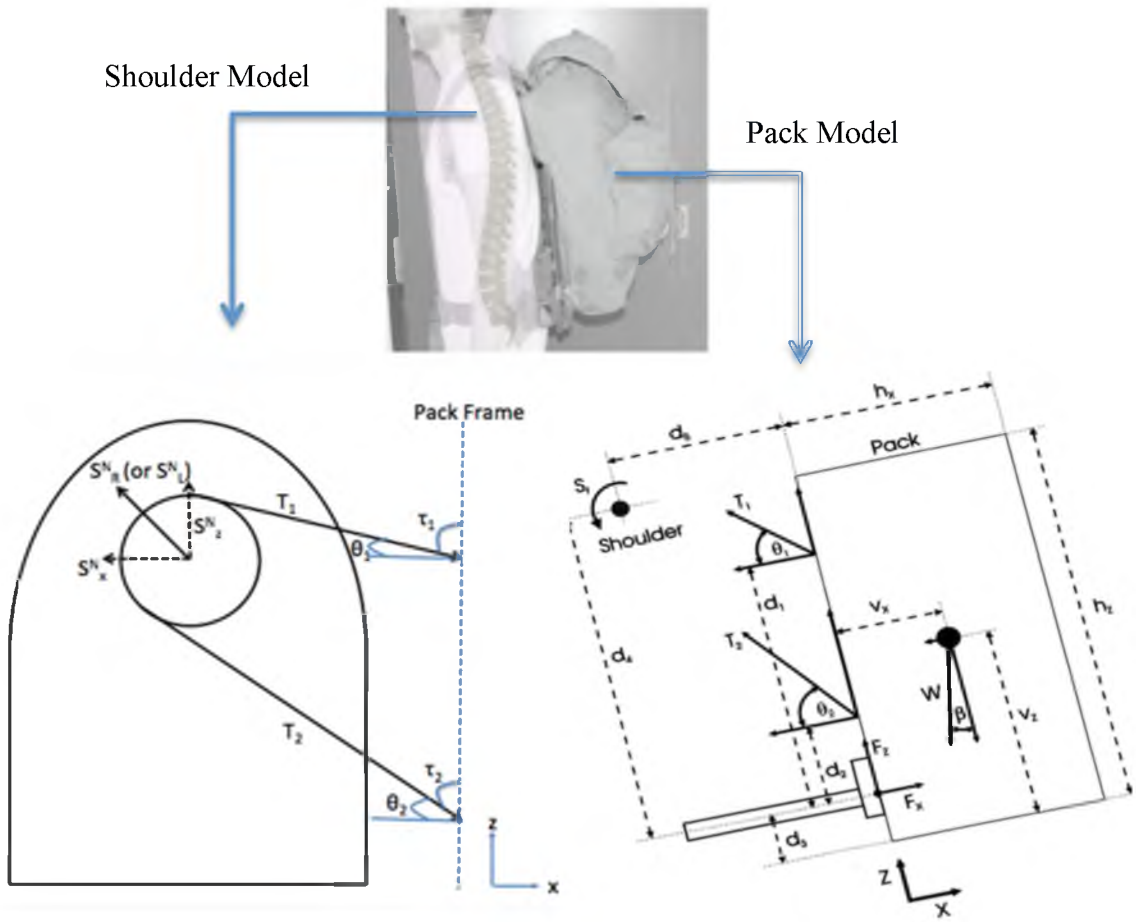

Figure 5-6. Biomechanical Model

Note: Models modified from (Pelot et al., 1998)

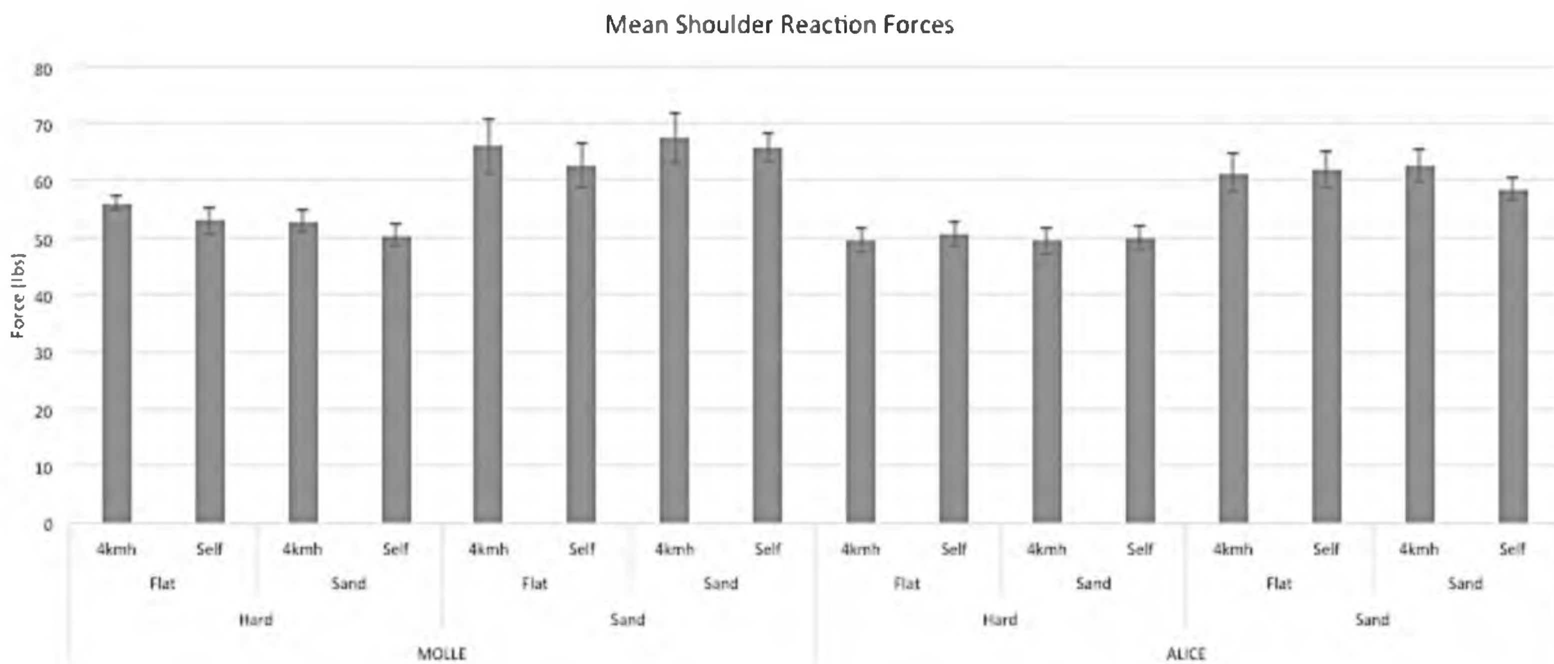

Figure 5-7. Mean Shoulder Reaction Force

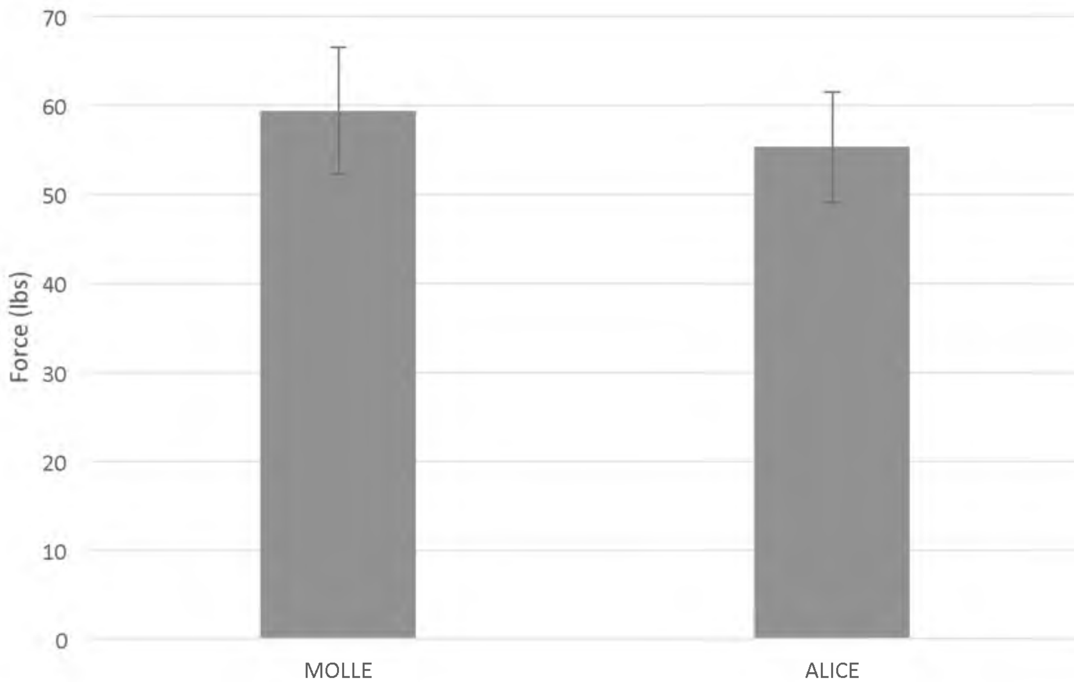

Figure 5-8. Shoulder Reaction Force by Backpack (p<.001)

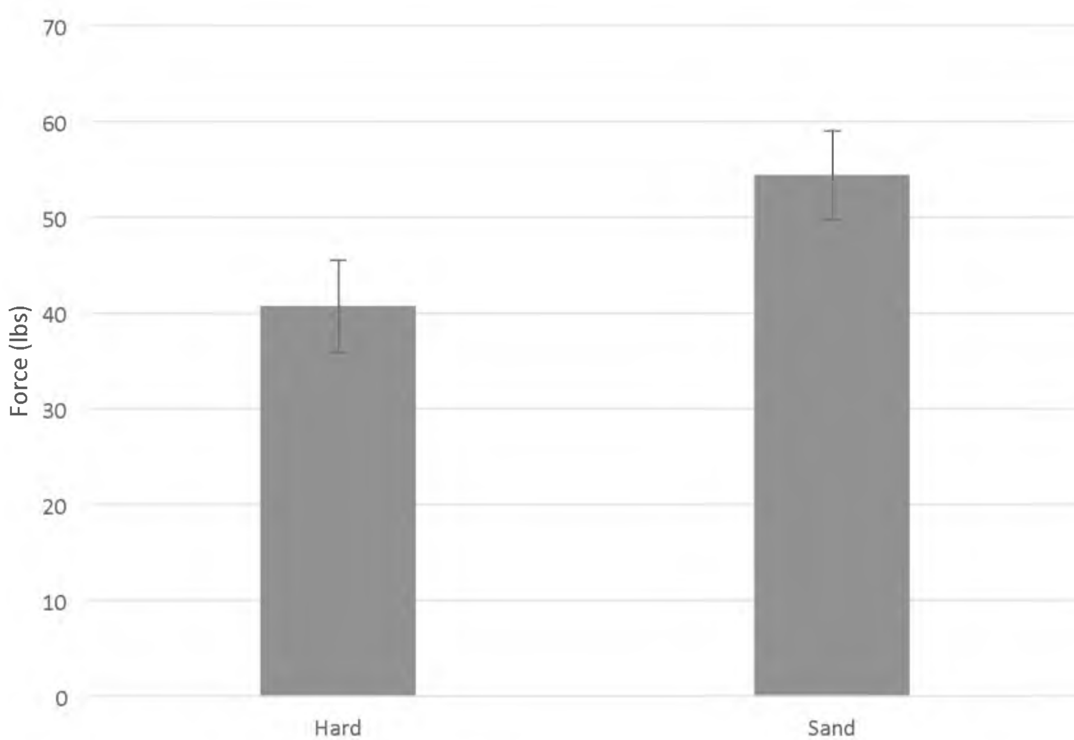

Figure 5-9. Shoulder Reaction Force by Surface (p<.001)

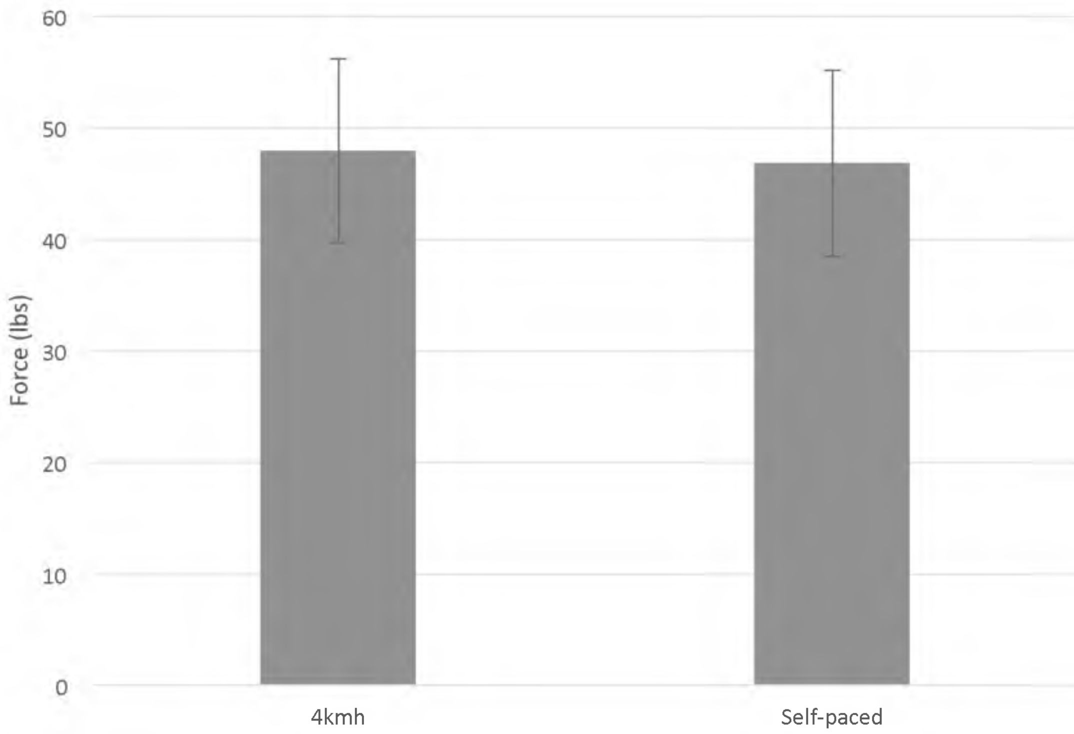

Figure 5-10. Shoulder Reaction Force by Speed (p=.003)

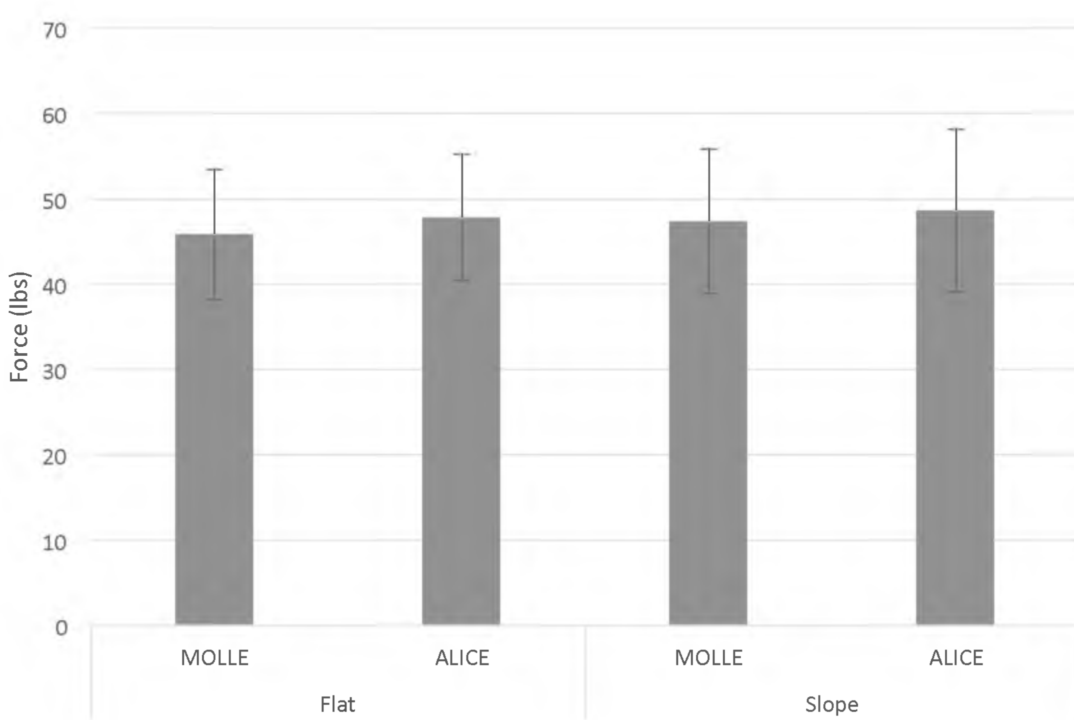

Figure 5-11. Shoulder Reaction Force by Slope and Backpack (p=.019)

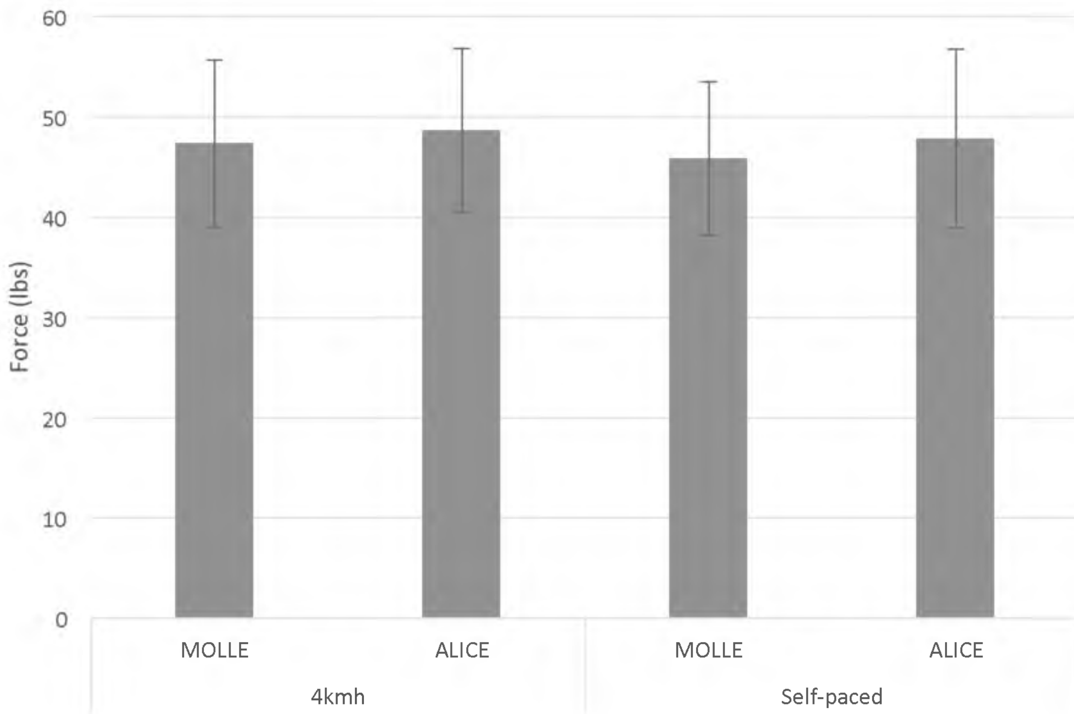

Figure 5-12. Shoulder Reaction Force by Speed and Backpack (p=.048)

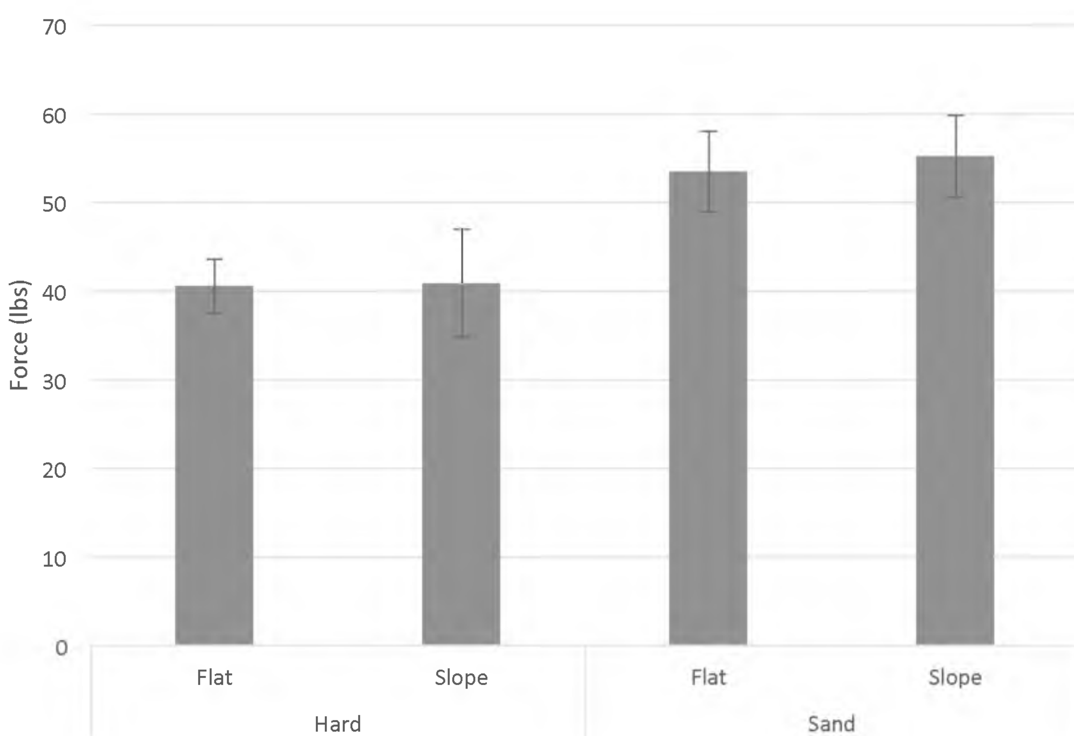

Figure 5-13. Shoulder Reaction Force by Surface and Slope (p=.005)

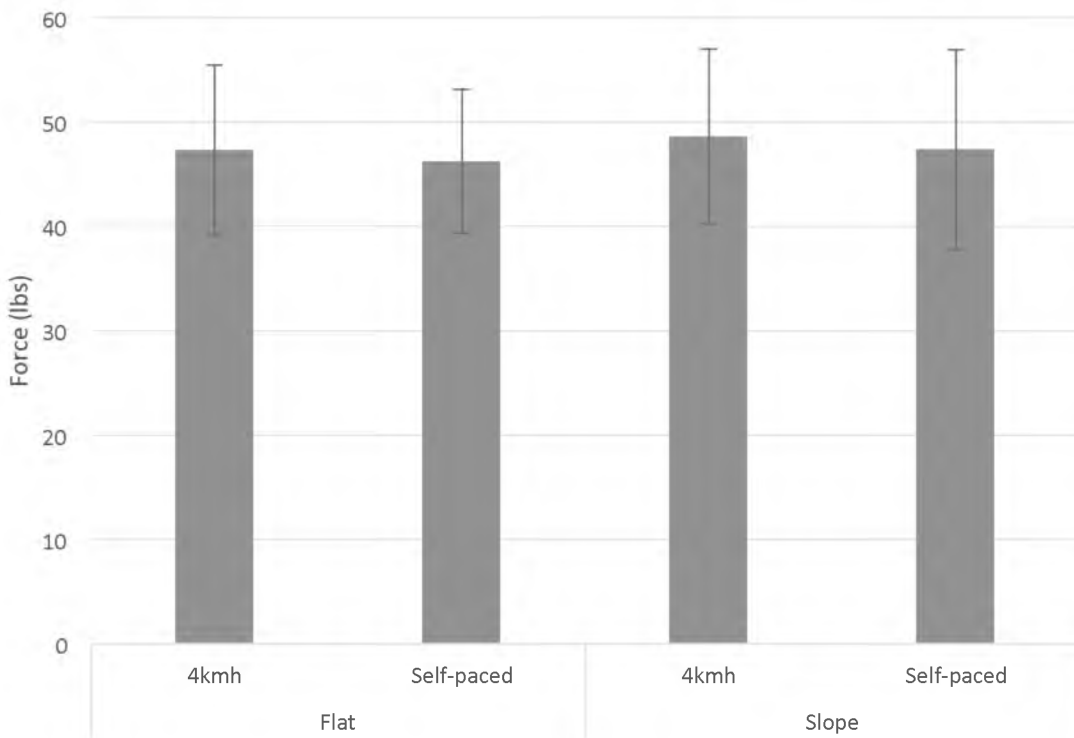

Figure 5-14. Shoulder Reaction Force by Slope and Speed (p=.012)

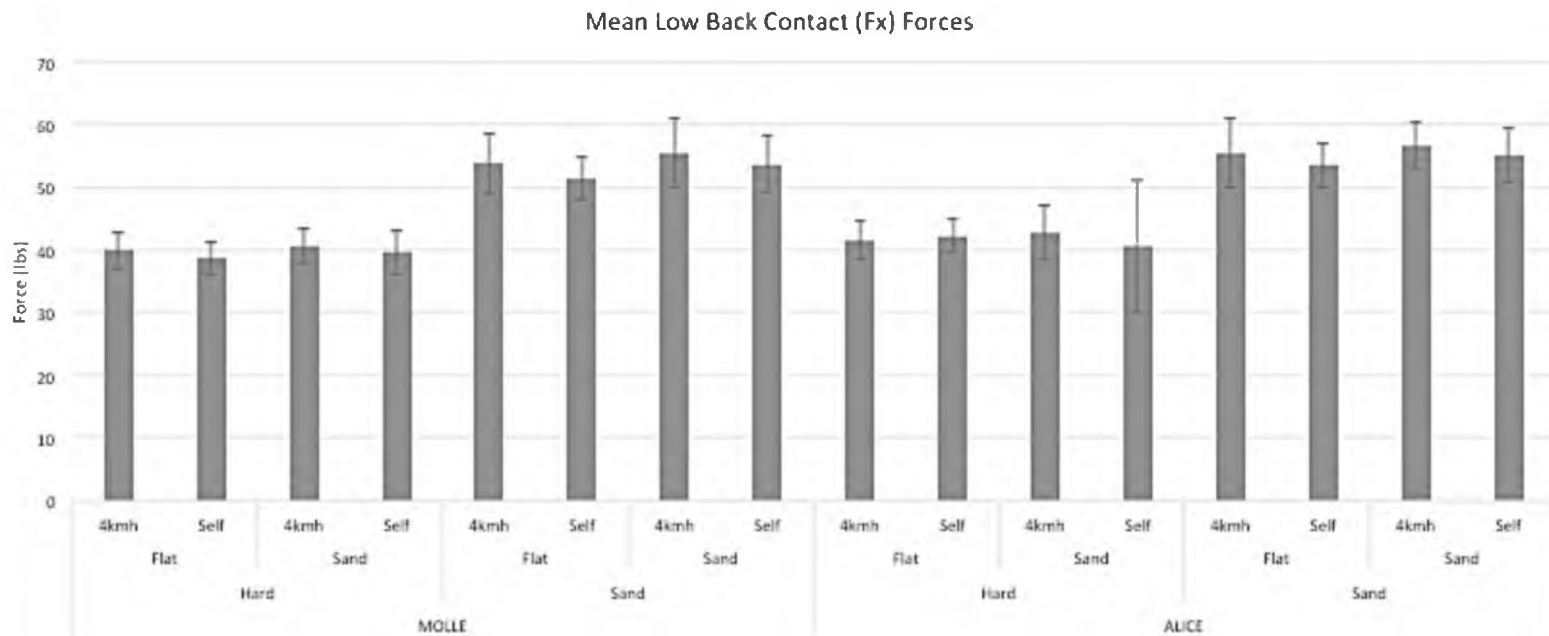

Figure 5-15. Mean Low Back Contact Force (*Fx*)

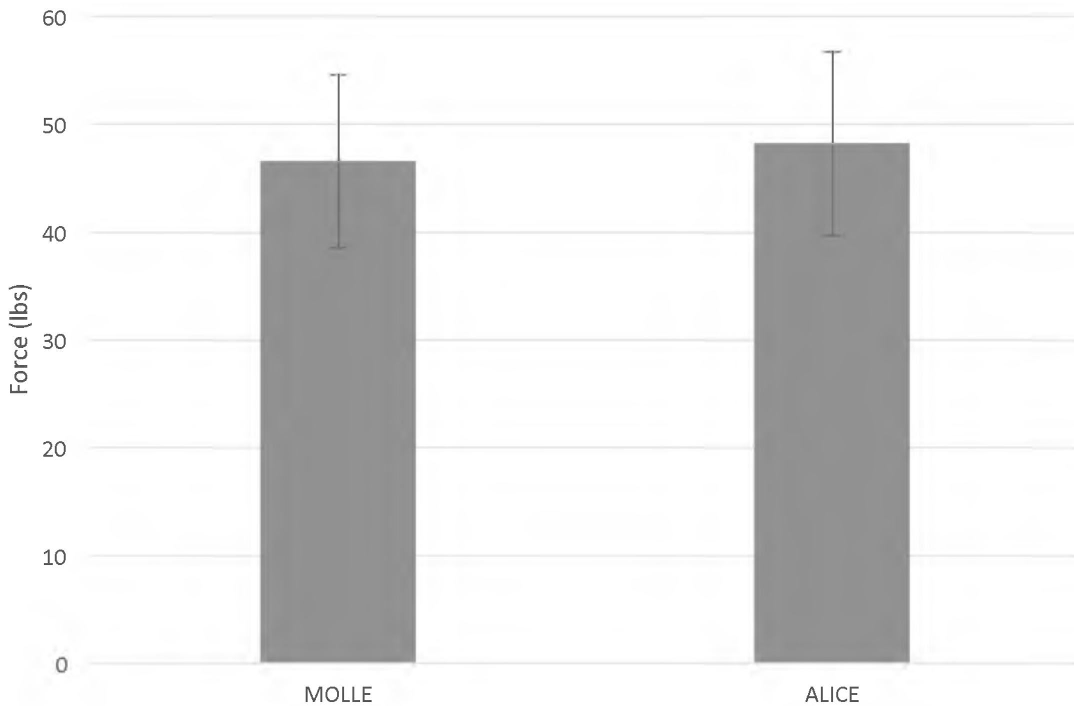

Figure 5-16. Low Back Contact Force by Backpack (*p*<.001)

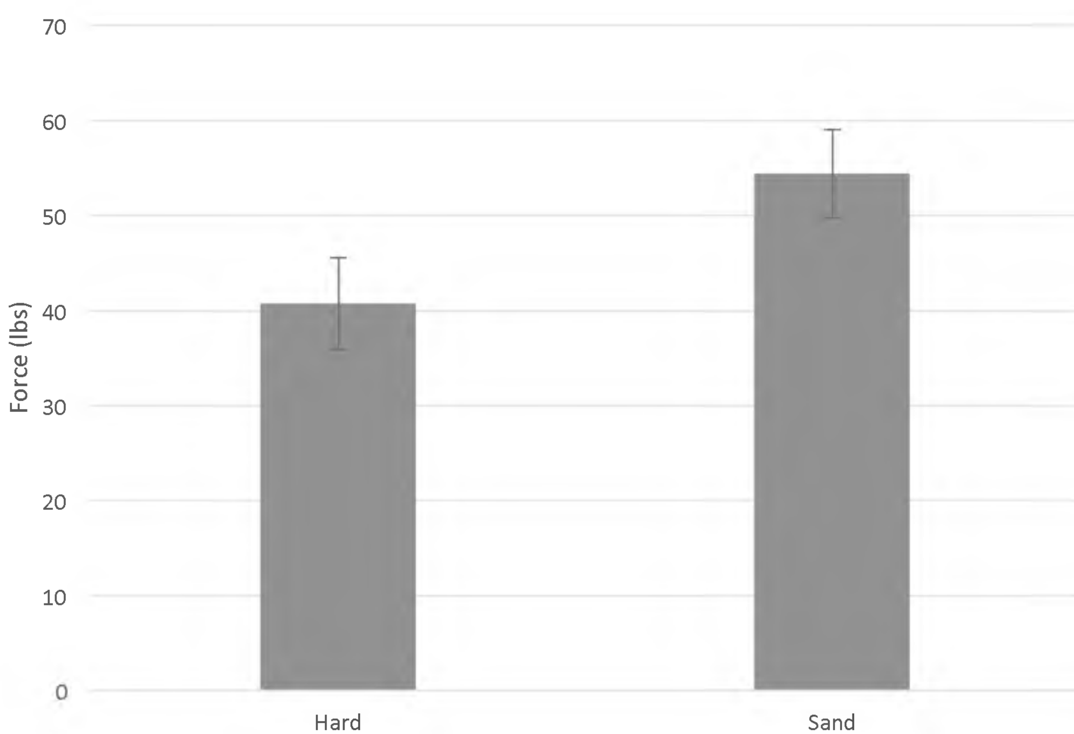

Figure 5-17. Low Back Contact Force by Surface (*p*<.001)

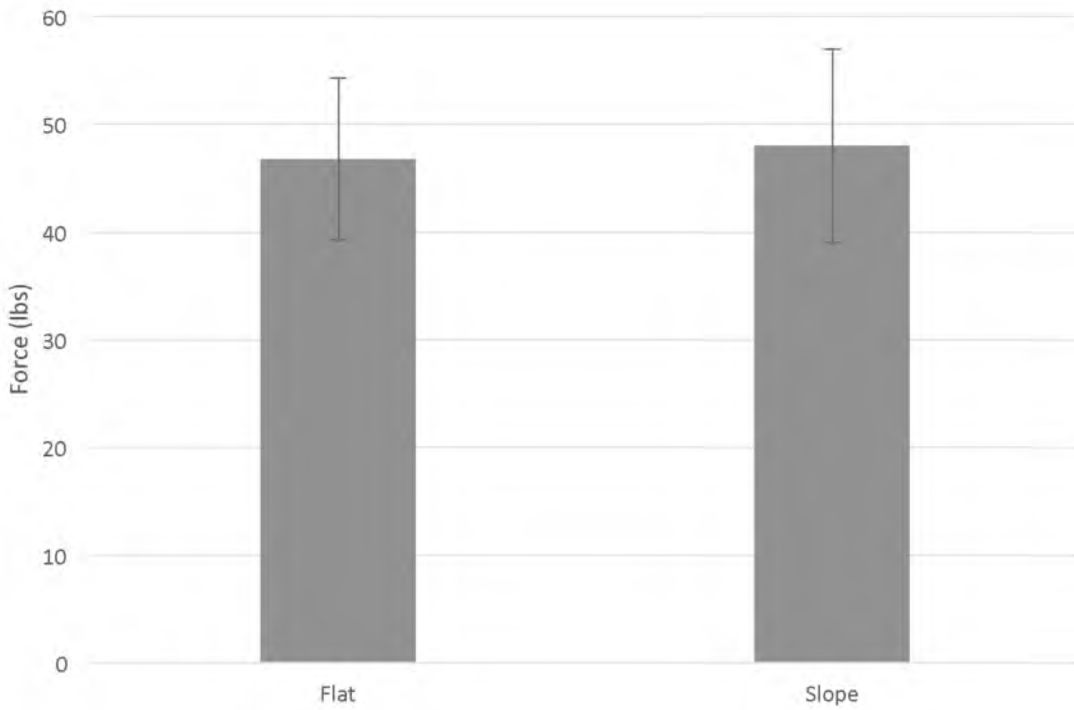

Figure 5-18. Low Back Contact Force by Slope (p=.005)

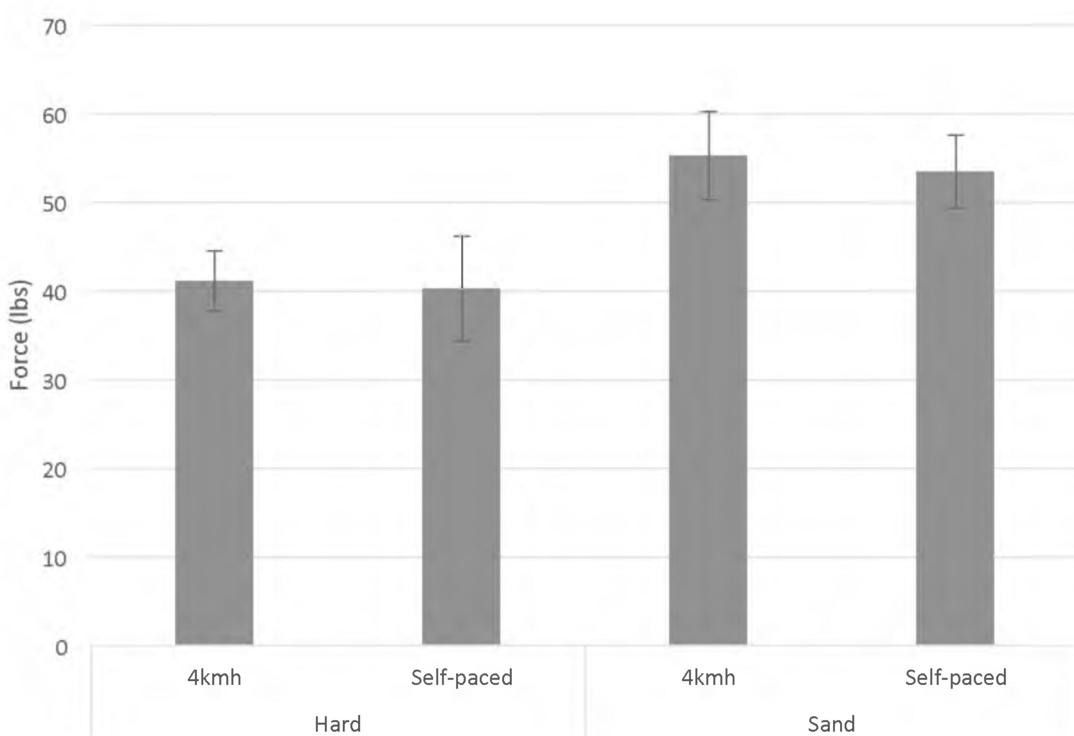

Figure 5-19. Low Back Contact Force by Surface and Speed (p=.009)

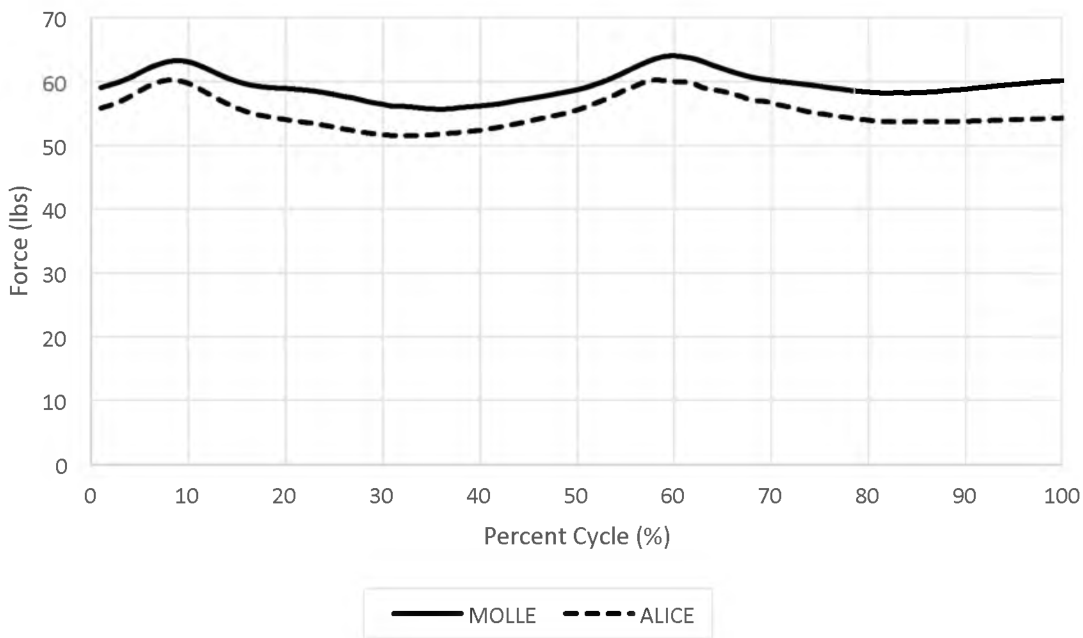

Figure 5-20. Shoulder Reaction Force Profile

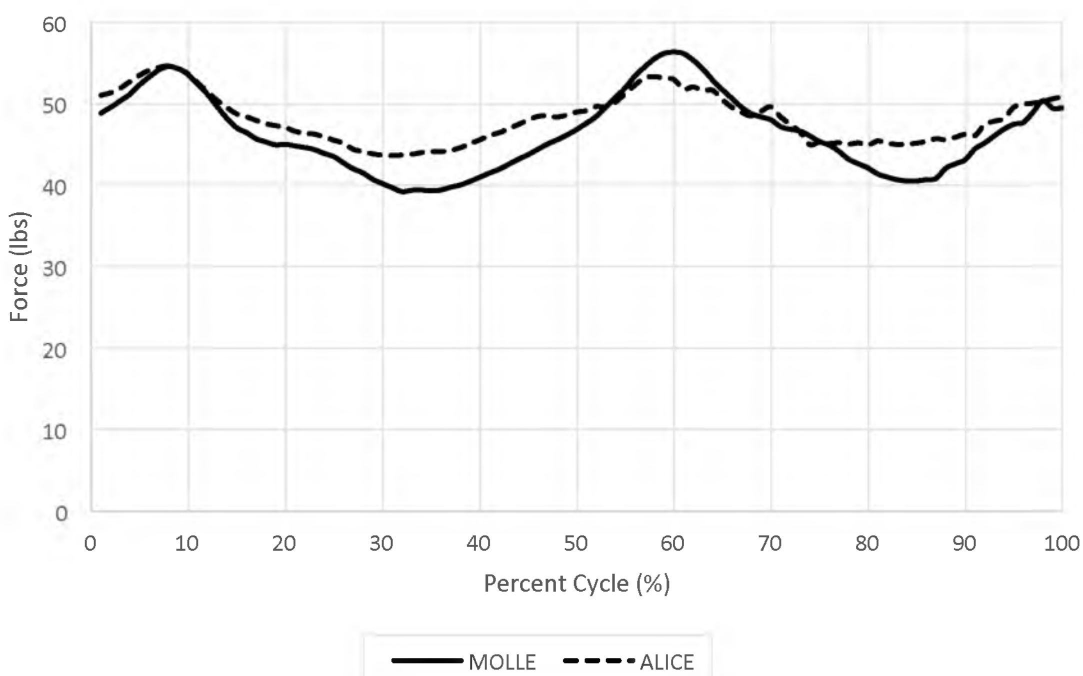

Figure 5-21. Low Back Contact Force Profile

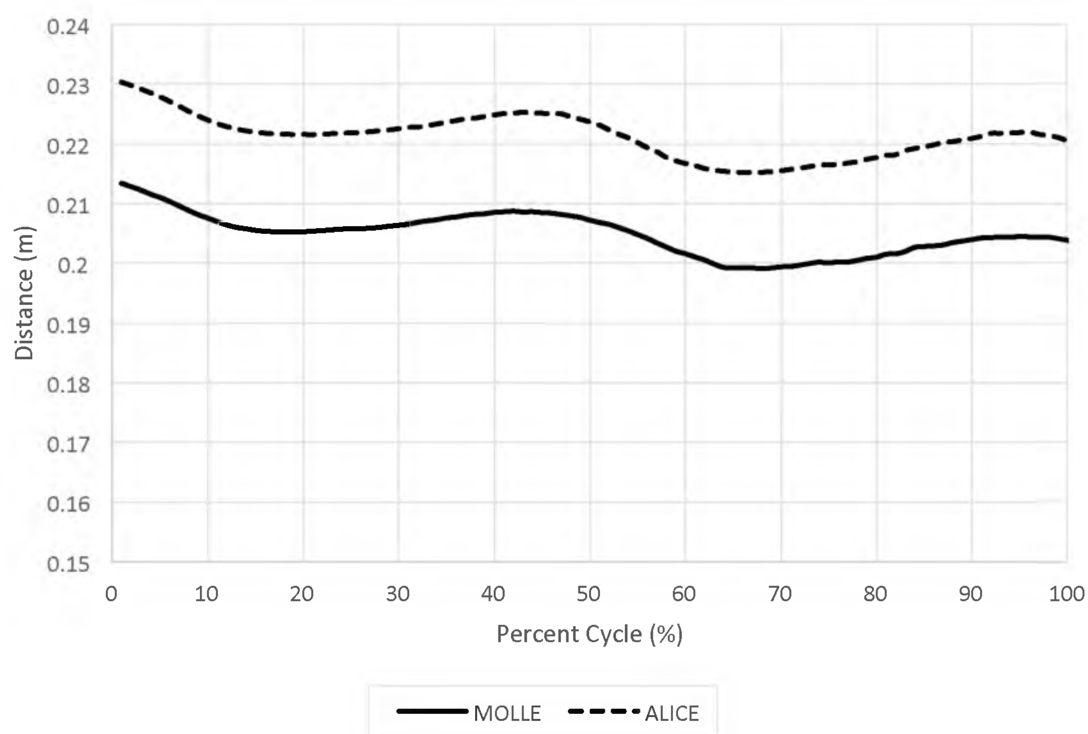

Figure 5-22. COM of Backpack Movement Profile (Sagittal)

Table 5-2. Evaluation by Pain Perception Level

	Shoulder	Low Back	Decision
Criteria (Pelot, 1998)	289 N	135 N	
MOLLE	262 N	204 N	Not Acceptable
ALICE	245 N	214 N	Not Acceptable

A

B

C

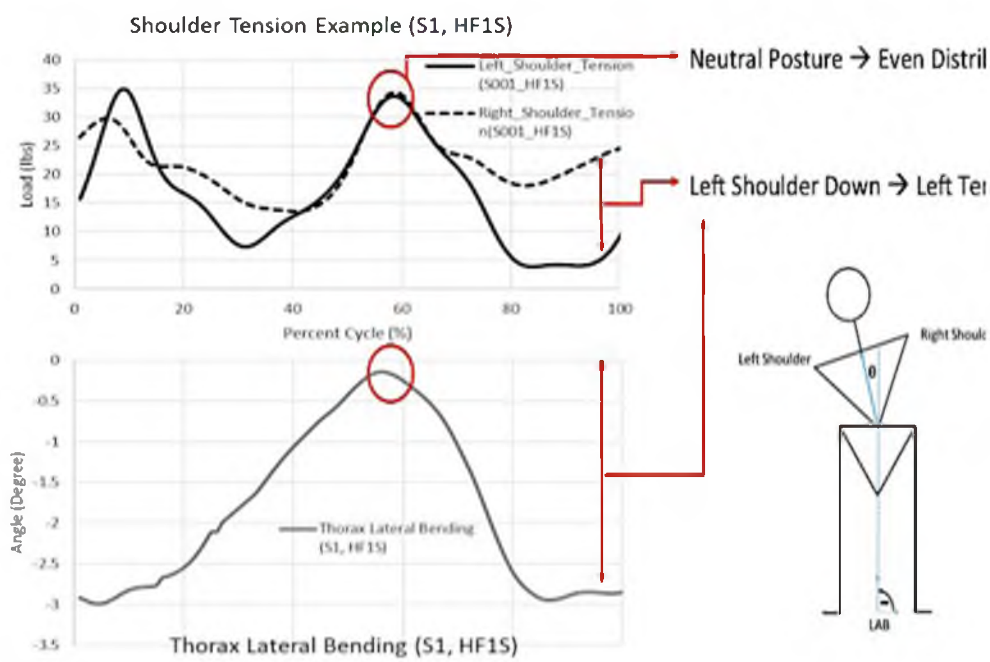

Figure 5-23. Relationship Between Shoulder Reaction Force Profile and Thoracic Lateral Bending.
A) Shoulder Tension; B) Thorax Lateral Bending; C) Upper Body Bending

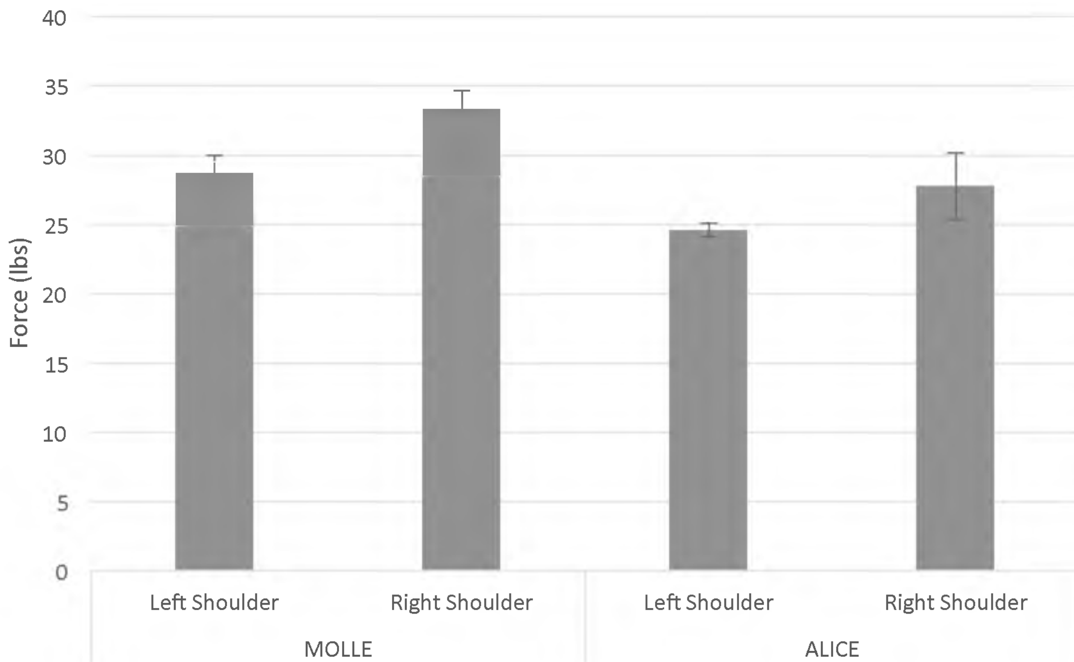

Figure 5-24. Shoulder Reaction Force Comparison (Left vs. Right)

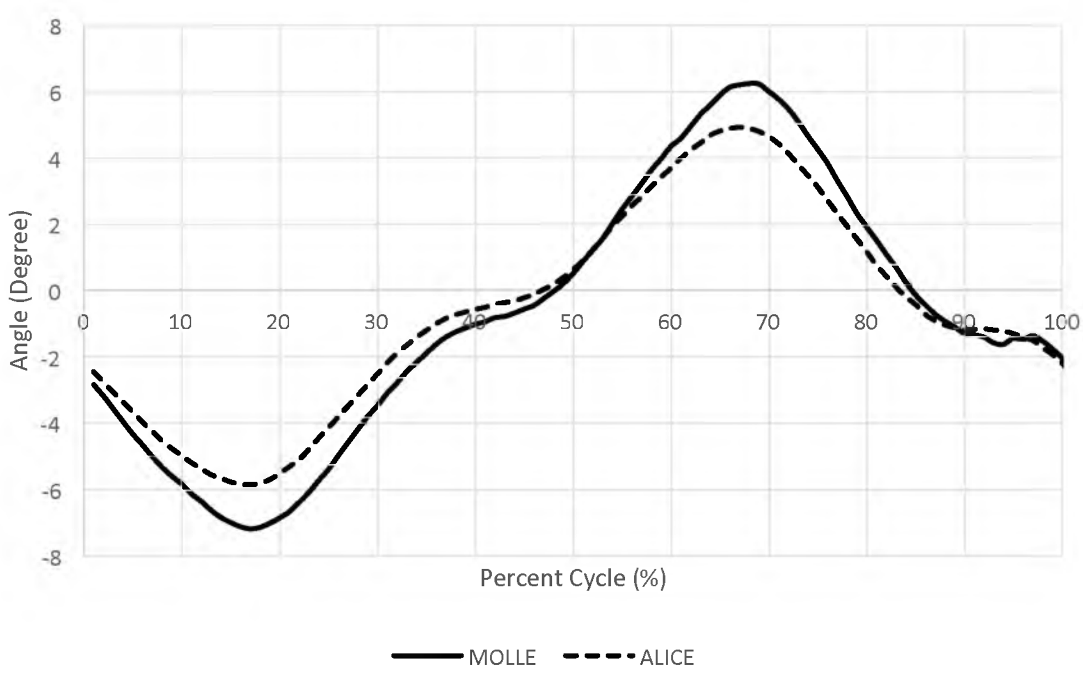

Figure 5-25. Thorax-Pelvis Coronal Bending Profile by Backpack

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CHAPTER 6

BIOMECHANICAL ESTIMATION OF FORCES ON L5/S1 LUMBOSACRAL DISC AND ITS IMPLICATION

The purpose of this study is to estimate the compressive and shear forces acting on the L5/S1 disc. Erector spinae muscle force, shoulder reaction force, upper body weight, and L5/S1 disc angle were used for the force calculation from the proposed biomechanical model. L5/S1 disc angle was derived from the relative motion between the thorax and pelvis. A greater compressive force was found on the sand surface than on the hard surface, but the forces did not exceed the NIOSH (The National Institute for Occupational Safety and Health) guidelines. Backpack load induced a greater shear force on L5/S1 disc significantly.

Introduction

It is well known that heavy load carriage is one of the main risk factors for musculoskeletal injuries in the military. Backpacks are a major load carriage method in the military and soldier loads are increasing significantly. Low back pain and injury are the most frequent medical cases in military operations and training (Hauret et al., 2010; Knapik et al., 1991). What is unclear, however, is the cause of the problem and the level of risk (Hoogendoorn et al., 1999; Knapik et al., 1996; Pope et al., 1998; Knapik et al., 2004).

Heavy load carriage is a risk factor for lower back injuries (Reynolds et al., 1990). Back pain in young people has been found to be related to heavily loaded backpacks (Negrini et al., 2002). Framed packs exert a consistent anterior force on the lower back, and it has been suggested that this force could contribute to low back pain and soreness (Lafiandra et al., 2004).

Biomechanical studies on loads at the lumbar spine have been carried out in attempts to establish a relationship between spine loads and lower back injuries. There is increasing evidence to indicate that such a relationship does exist.

Khoo et al. proposed the biomechanical model for estimating peak lumbosacral forces while walking. According to his methods, back compressive force can be estimated using an inverse dynamic method (1994). Goh analyzed effects of backpack loads on peak forces in the lumbosacral spine using the model of Khoo et al. They found that increases in loads significantly increase mean L5/S1 forces. They also found that the compression force component was more dominant than the sheer force component on the L5/S1disc (Goh et al., 1998).

Method

Biomechanical Model

For indirect estimation of forces (compressive force: F_c, shear force: F_s) on L5/S1 disc, an L5/S1 model is proposed (Figure 6-1). To build the biomechanical L5/S1 model, some assumptions have been made:

- A1. Shoulder reaction force (S^N) and low back contact force (F_x): Reaction forces existing on the shoulders and hips can be calculated using the load cells attached on each shoulder strap. Force on the hip area (F_x), however, does not affect the forces on L5/S1 because the hip belt is located on the pelvis. Thus, it can be negligible.

- A2. Erector spinae muscle force (F_{ES}): Using erector spinae EMG activity and a previously determined force-EMG calibration relationship for each subject, erector spinae muscle force will be estimated during the gait cycle. Additionally, the line of action of F_{ES} is parallel to the F_c.
- A3. Upper body weight is 50 % of whole body weight and acts along the gravitational direction (z-direction from the lab coordinate system).

From the assumptions above, three main force components are acting on L5/S1 lumbar spine segment: shoulder reaction force (S^N), upper body weight (UBW), and erector spinae muscle force (F_{ES}).

In Figure 6-1, α represents the angle between L5/S1 and the horizontal. For static estimation, Chaffin derived the angle from the relationship between knee flexion angle and upper body flexion angle (Andersson, 1991). We estimated the angle from the relative thorax-pelvis flexion angle in the sagittal plane (γ) using the following equation.

$$\alpha = 40 + \gamma \text{ (Degree)}$$

Thus, the compressive force on L5/S1 (F_c) can be calculated from the equation:

$$F_c = \left(S_z^{N'} + UBW\right) \cos\alpha + F_{ES}$$

$S_z^{N'}$ represents the vertical component (z-axis) of the shoulder reaction force with regard to our lab coordinate system. The shear force on L5/S1 (F_s) can be estimated from the following equation.

$$F_s = \left(S_z^{N'} + UBW\right)\sin\alpha$$

Data Analysis

The collected data were analysed using SPSS software (Ver.18.0, IBM Corporation, Armonk, NY) at a significance level of 0.05. Compressive force and shear force on the L5/S1 disc were analyzed using MANOVA. Gait cycle was normalized from 0 % (initial left heel strike) to 100 % (secondary left heel strike).

Results

Compressive Force on L5/S1 Disc

Figure 6-2 shows the average compressive force acting on the L5/S1 spinal disc. After analysis with a MANOVA test, the back compressive forces showed statistically significant differences between surfaces and load scenarios. A higher force was found on sand than on the hard surface (p<.001; Figure 6-3). Interaction between slope (no slope, slope) and backpack load and type (no load, MOLLE, ALICE; p=.48; Figure 6-4) was also significant.

Shear Force on L5/S1 Disc

Figure 6-5 shows the average shear force on the L5/S1 disc. From the MANOVA analysis, the lumbosacral shear forces showed statistically significant differences. A higher force was found on sand than on the hard surface (p<.001; Figure 6-6); on the flat surface than on the sloped surface (p<.001; Figure 6-7); at 4 km/h than at self-paced speed (p<.001; Figure 6-8); and when walking with a backpack load than with no load (p<.001; Figure 6-9).

Additionally, interactions between slope (no slope, slope) and backpack load and type (no load, MOLLE, ALICE; p<.001; Figure 6-10); surface type (hard, sand) and backpack (p<.001; Figure 6-11); speed (4 km/h, self-paced) and backpack (p<.001; Figure 6-12); surface type and slope (p=.003; Figure 6-13); and surface type and speed (p=.005; Figure 6-14) were statistically significant.

Discussion

Estimation of L5/S1 Disc Angular Displacement Profile

Estimating the angular change of the L5/S1 disc is essential for the calculation of forces on the disc. Chaffin et al. derived the angle from the relationship between the knee flexion angle and the trunk flexion angle in a static posture (1991). Gilliam et al. insisted that movement of the sacrum is proportional to the pelvic movement (1994). During et al. addressed that there is a clear biomechanical relationship between the pelvic and the sacral angle, and therefore the pelvic angle changes in accordance with the sacral angle (1985). Day et al. reported a correlation between the pelvic tilt and the lumbar curve (1984).

No effective method, however, has been proposed for a dynamic estimation of the angular change of the disc. In our study, we estimated the angle from the relative angle between the thoracic sagittal flexion and the pelvic tilt angle.

Figure 6-15 shows the calculated average L5/S1 disc angles from the proposed method for each backpack load and type. When there was no load during gait, the mean sacral angle was 43.6 degrees below the horizontal. It is known that 40 to 44 degrees are the average sacral angles for normal subjects (Gilliam et al., 1994), and our method also showed reasonable sacral angles for normal walking conditions. As seen in Chapter 3, there was a significant increase in the thoracic flexion angle with a backpack carriage. The pelvic tilt angle was also increased due to the backpack load but the magnitude of the pelvic angular change was smaller than that of the thorax. This resulted in negative relative thorax-pelvic sagittal angles and smaller L5/S1 disc angles during backpack-loaded gait vs unloaded gait.

Figure 6-16 illustrates the angular displacement profiles by each loading condition. Peak angular displacement was identified at around 0 % to 10 % and 50 % to 60 % of the gait cycle, respectively.

The proposed calculation methods in our study can be a good starting point for a dynamic estimation of L5/S1 angle that can be applicable to gait studies. Additional research is required to validate the estimation method in the future.

Effects of Backpack Carriage on Forces on the L5/S1 Disc

Table 6-1 summarizes the post-hoc analysis (Tukey LSD) results to identify the detailed differences of each loading condition. The backpack was not a significant factor in the back compressive force, but showed a statistically significant difference in the back shear force. No difference was found between MOLLE and ALICE designs.

Figure 6-17 compares average compressive forces on the L5/S1 disc by loading conditions. There was about 10 lbs of increase in the back compressive force from normal walking due to backpack carriage. The differences in force, however, were not statistically significant, and the compressive forces fell within NIOSH guidelines. Even though it was not statistically significant and met the NIOSH criteria, we could not rule out the potential risks from prolonged exposure to backpack loading. This will be further discussed in Chapter 7.

The effect of the backpack showed statistical significance in the back shear force and there was approximately a 44.4 % increase in the shear force between unloaded walking conditions and loaded backpack carriage. No difference was identified between MOLLE and ALICE. Figure 6-18 shows the mean of the back shear forces by each loading condition.

As seen in Figure 6-19, a higher shear force fluctuation was identified when carrying a backpack as a result of the shoulder reaction force fluctuation. The fluctuation might result in microfracture of a spinal disc after a period of exposure time, but it remained unclear in our study since no effective guideline was available.

Conclusion

The back compressive forces were within NIOSH guidelines. Even though it did not have a statistical significance and met the NIOSH criteria, we could not rule out the potential risks from the prolonged exposure to backpack loading.

The shoulder forces due to a backpack load resulted in increased back compressive and shear forces on the L5/S1 disc with the combinations of upper body weight and erector spinae muscle force. The shoulder forces induced a higher shear force fluctuation when carrying a backpack. The fluctuation might result in a microfracture of a spinal disc after a period of exposure, but it remained unclear in our study since no effective guideline was available.

There were significant increases in compressive and shear lumbar forces on the sand surface. Increased peak compressive stresses and the number of cycles resulted in a dramatic increase of the disc failure probability on the sand surface.

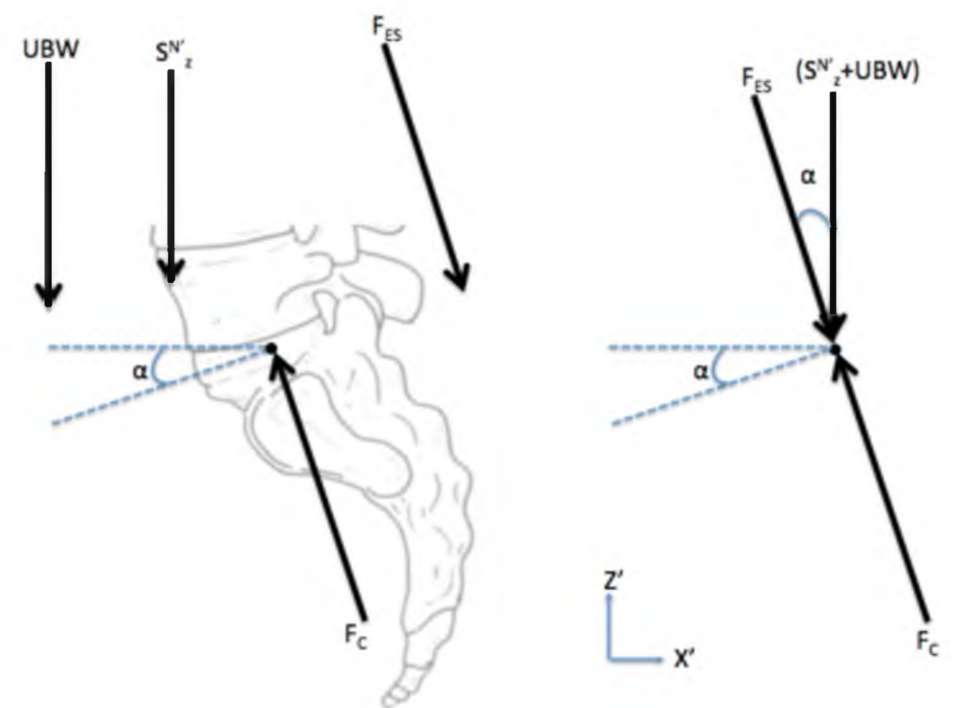

Figure 6-1. L5/S1 Model

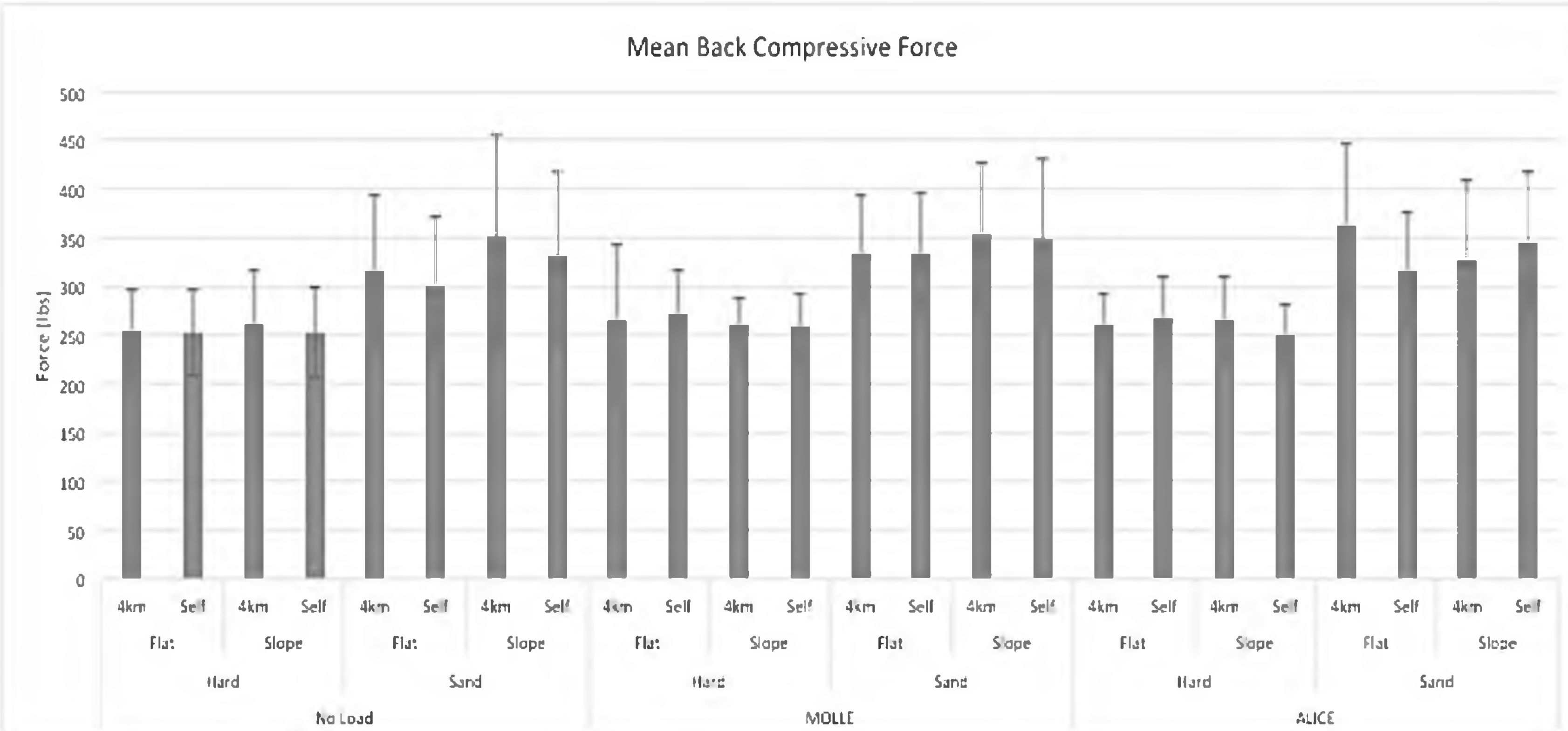

Figure 6-2. Mean Compressive Force on L5/S1

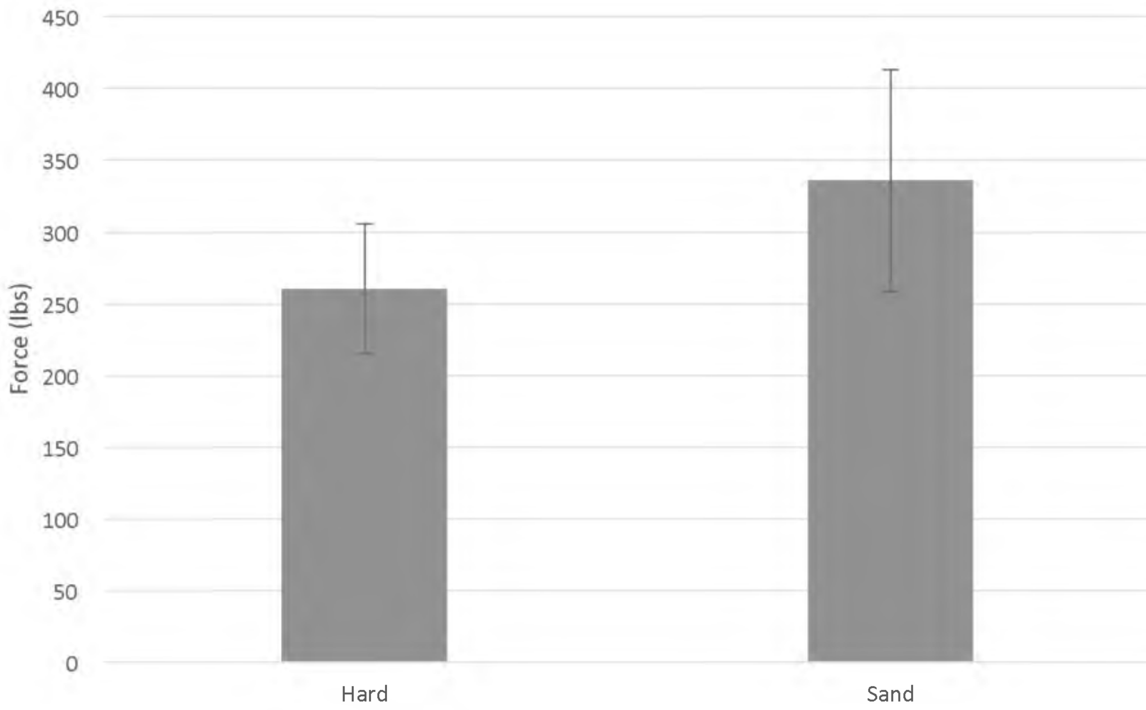

Figure 6-3. Compressive Force on L5/S1 by Surface (p<.001)

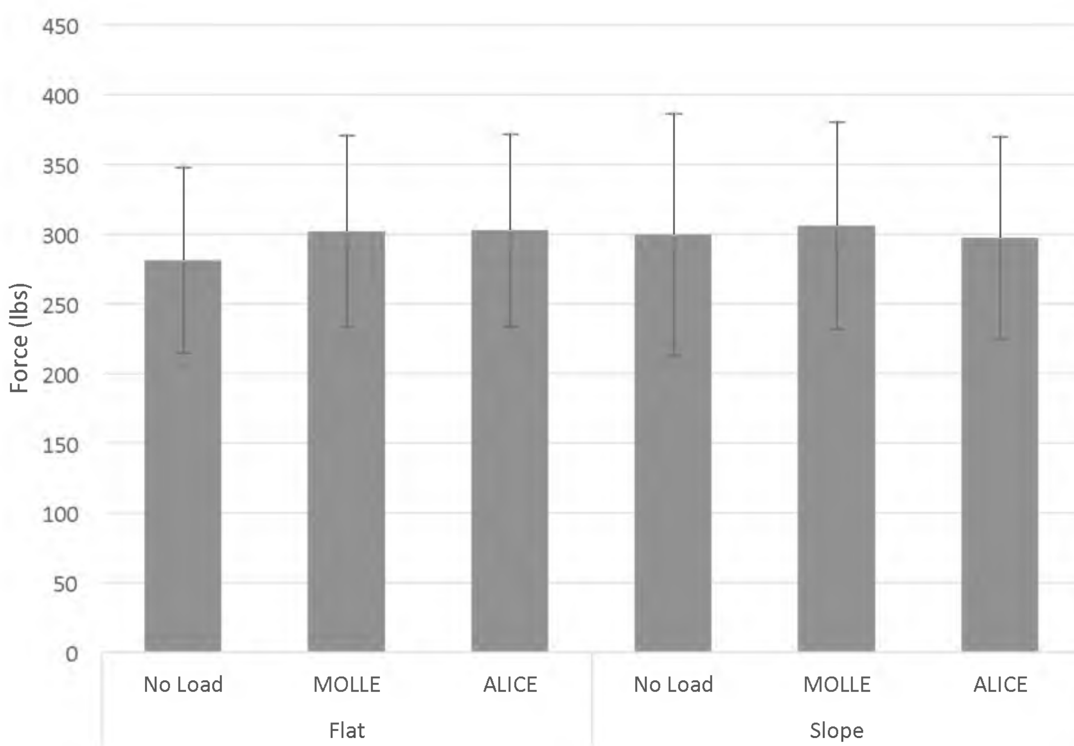

Figure 6-4. Compressive Force on L5/S1 by Slope and Backpack (p=.048)

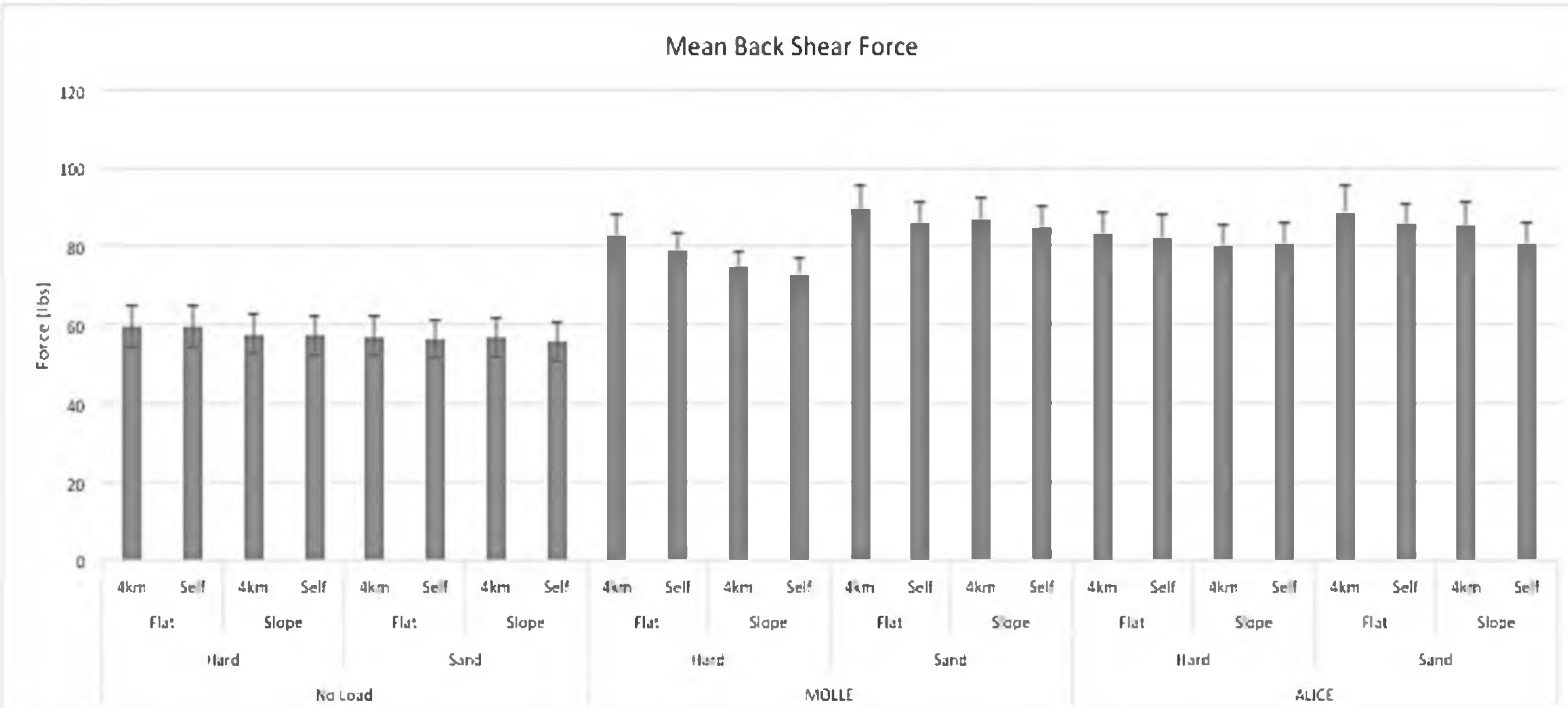

Figure 6-5. Mean Shear Force on L5/S1

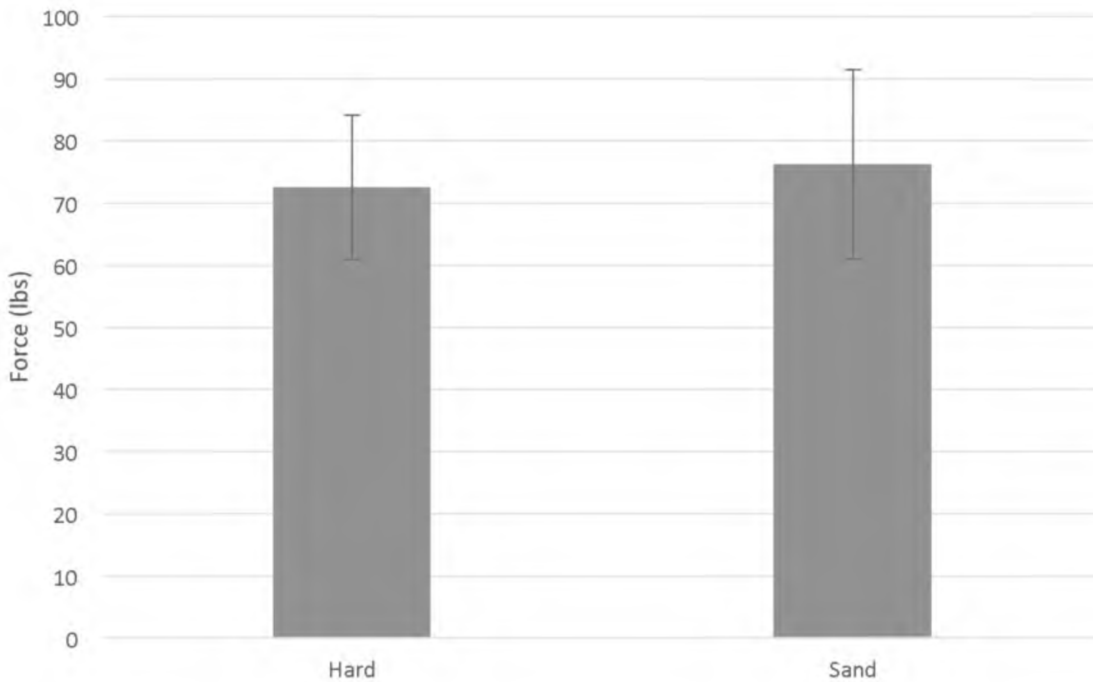

Figure 6-6. Shear Force on L5/S1 by Surface (p<.001)

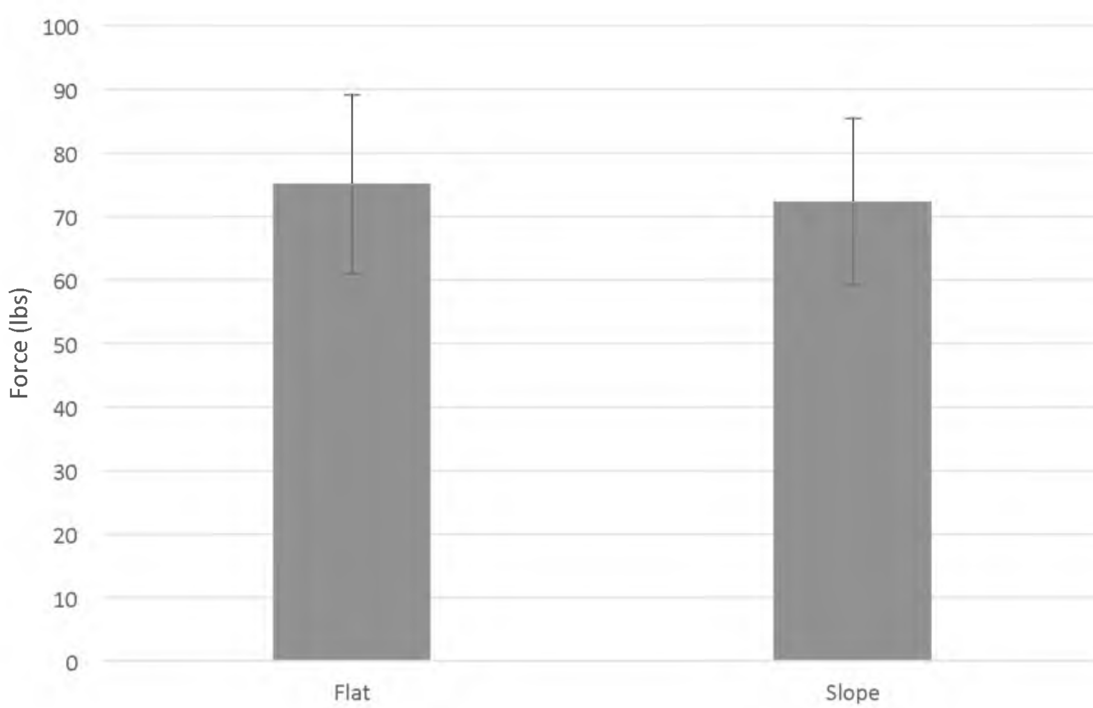

Figure 6-7. Shear Force on L5/S1 by Slope (p<.001)

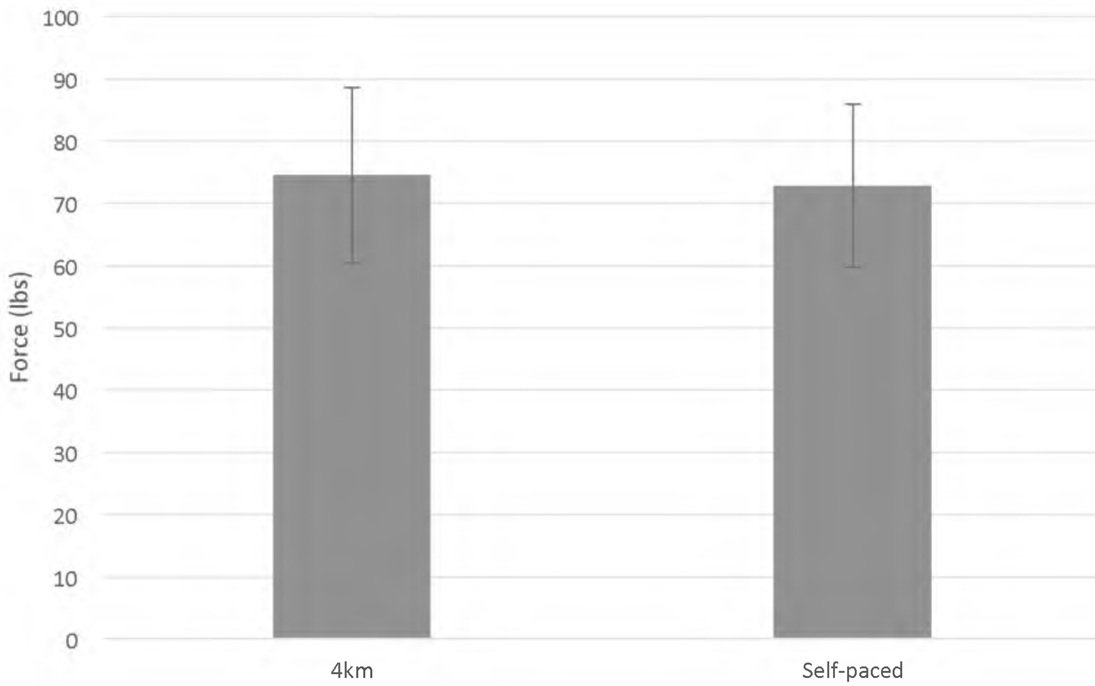

Figure 6-8. Shear Force on L5/S1 by Speed ($p<.001$)

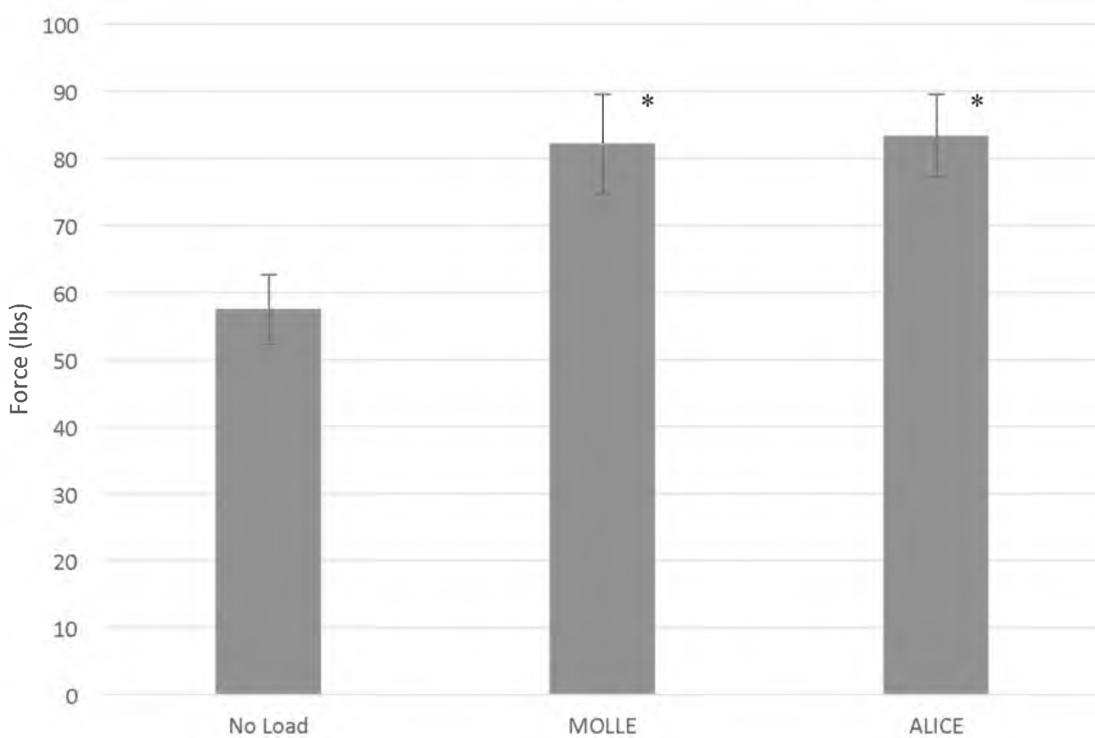

Figure 6-9. Shear Force on L5/S1 by Backpack ($p<.001$)
* indicates statistical significance ($p<.005$) compared to No Load

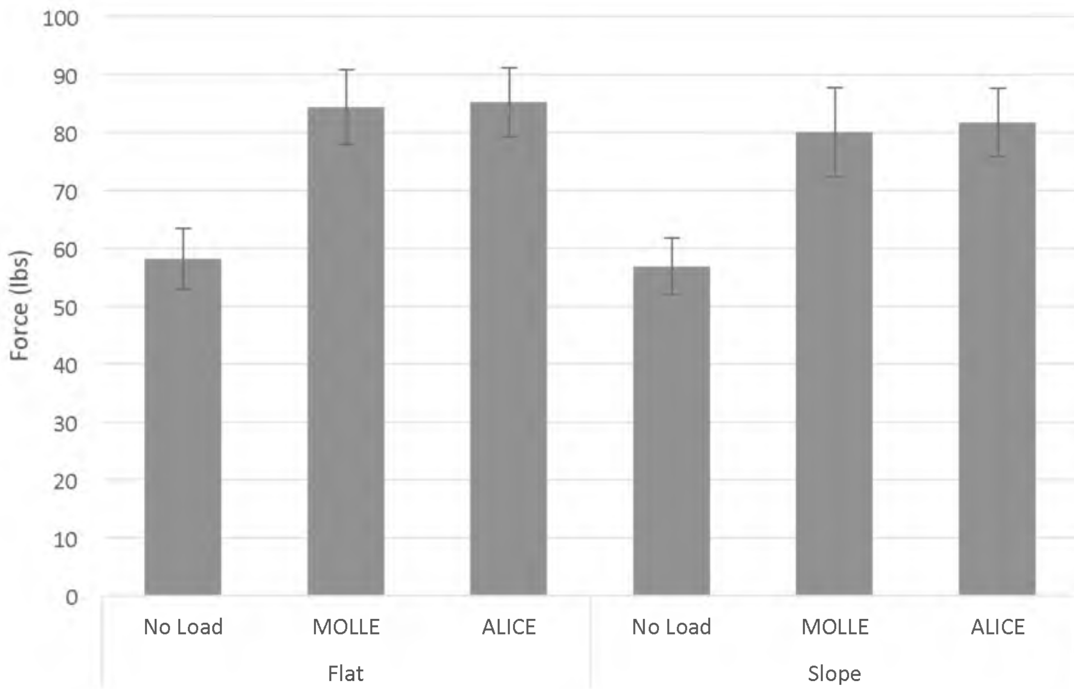

Figure 6-10. Shear Force on L5/S1 by Slope and Backpack ($p<.001$)

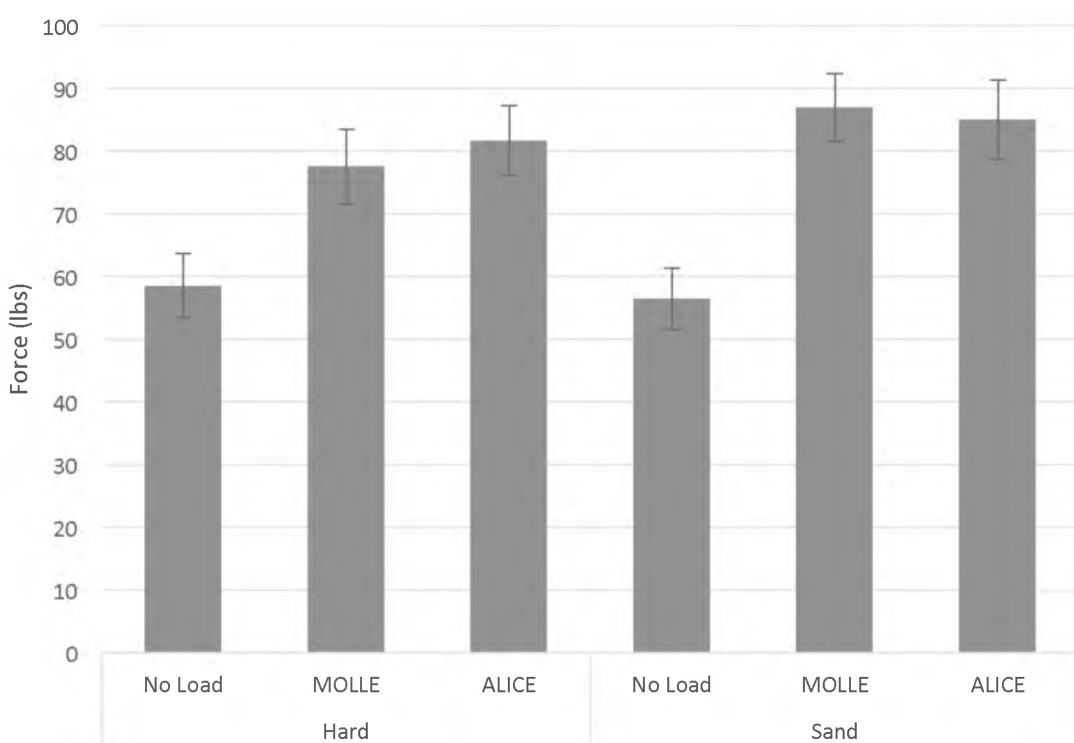

Figure 6-11. Shear Force on L5/S1 by Surface and Backpack ($p<.001$)

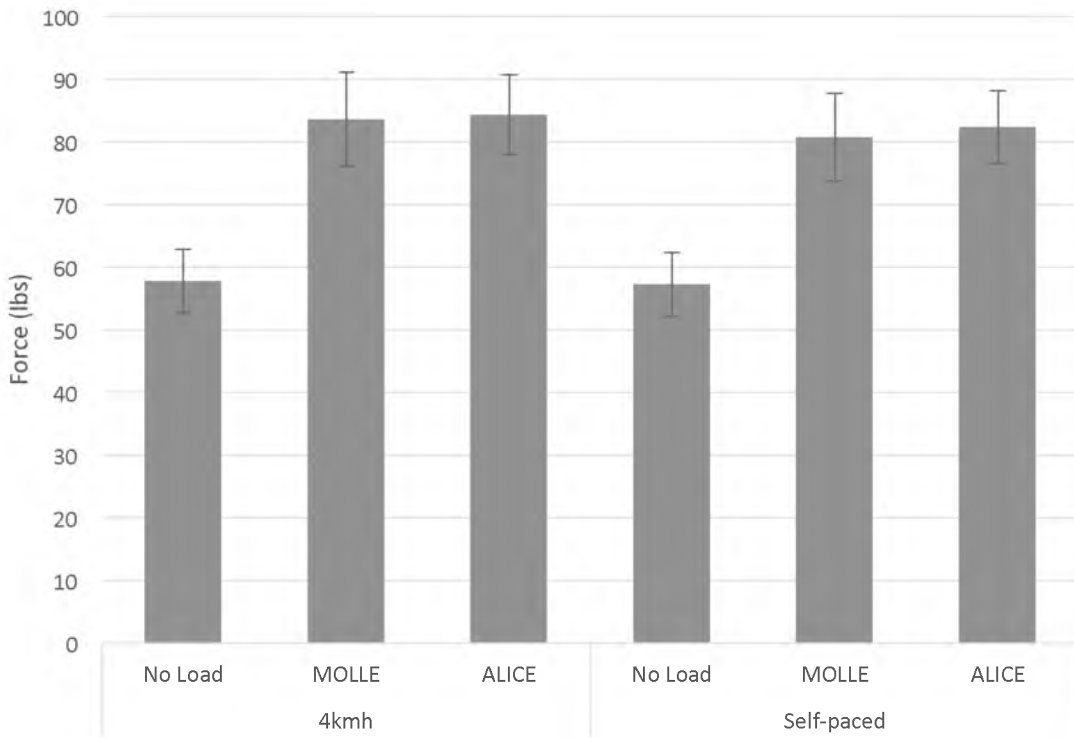

Figure 6-12. Shear Force on L5/S1 by Speed and Backpack (p<.001)

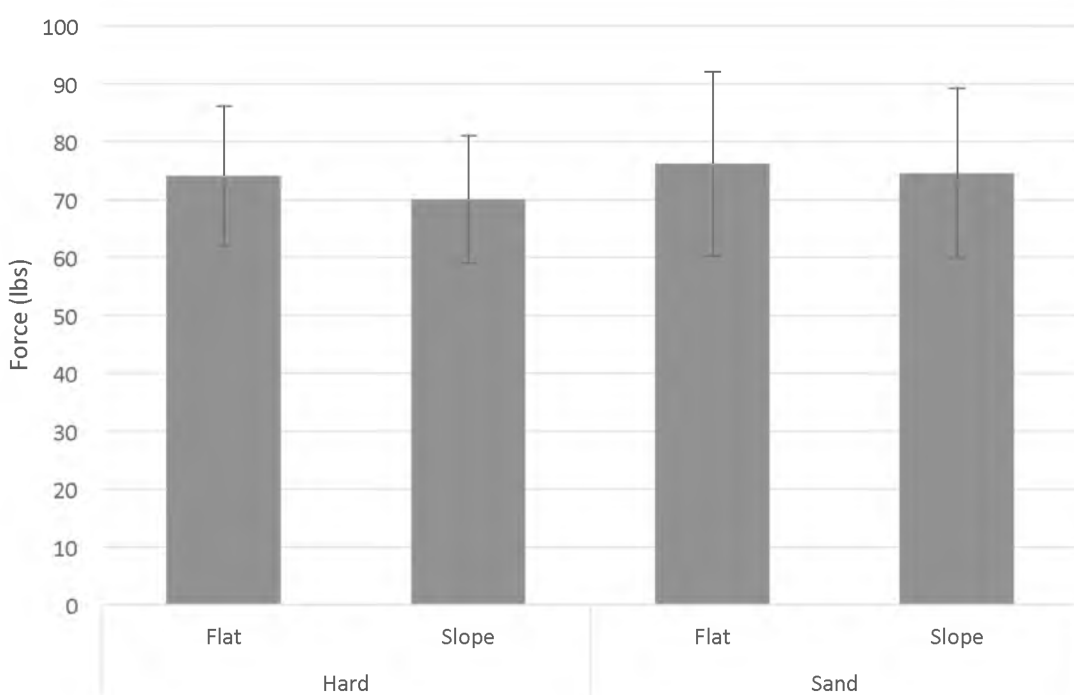

Figure 6-13. Shear Force on L5/S1 by Surface and Slope (p=.003)

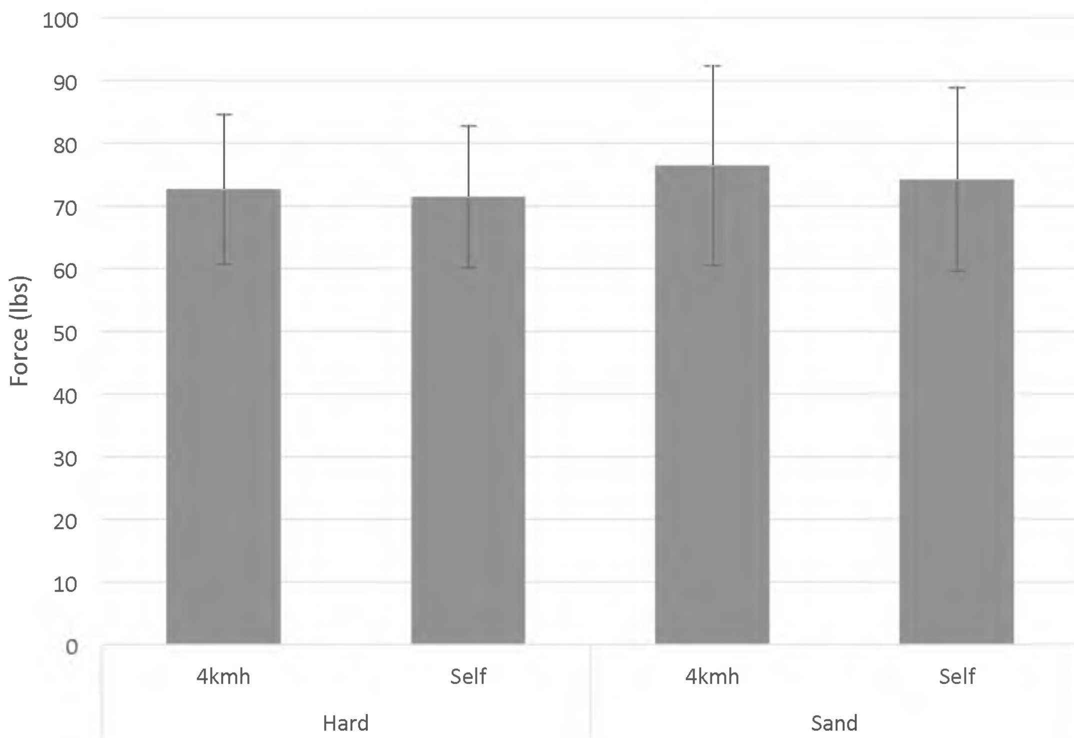

Figure 6-14. Shear Force on L5/S1 by Surface and Speed (p=.005)

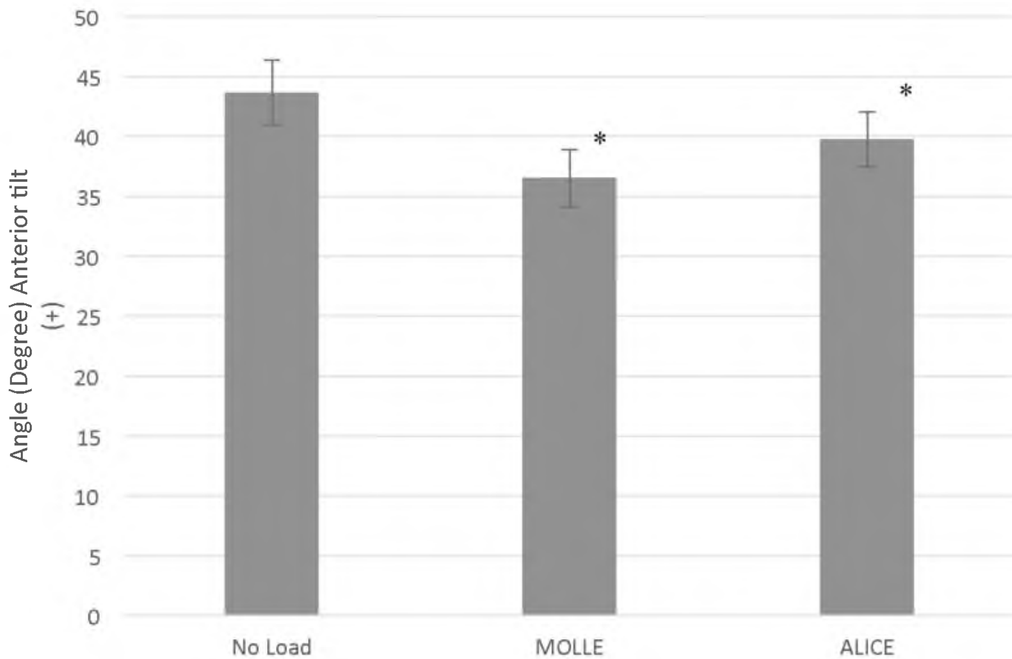

Figure 6-15. L5/S1 Angle
* indicates statistical significance ($p<.005$) compared to No Load

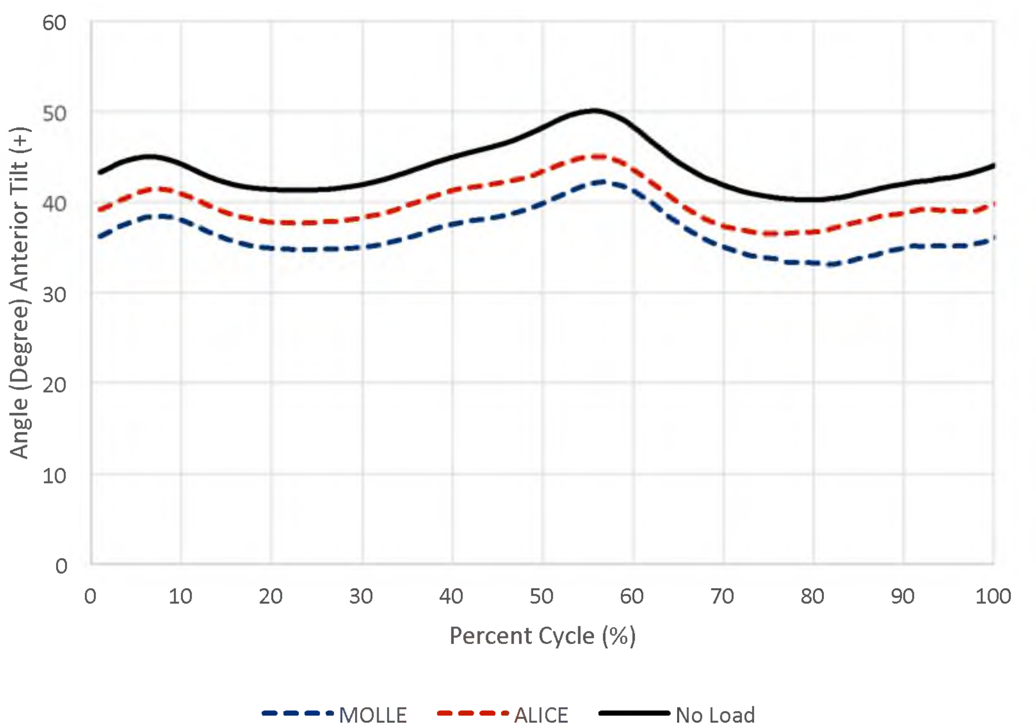

Figure 6-16. L5/S1 Angular Displacement Profile

Table 6-1. Effect of Backpack Carriage Forces on L5/S1 Disc

Post hoc			
	No Load vs. ALICE	No Load vs. ALICE	MOLLE vs. ALICE
Compressive	NA	NA	NA
Shear	.000	.000	.636

Note: Numbers in the table represent *p* values.

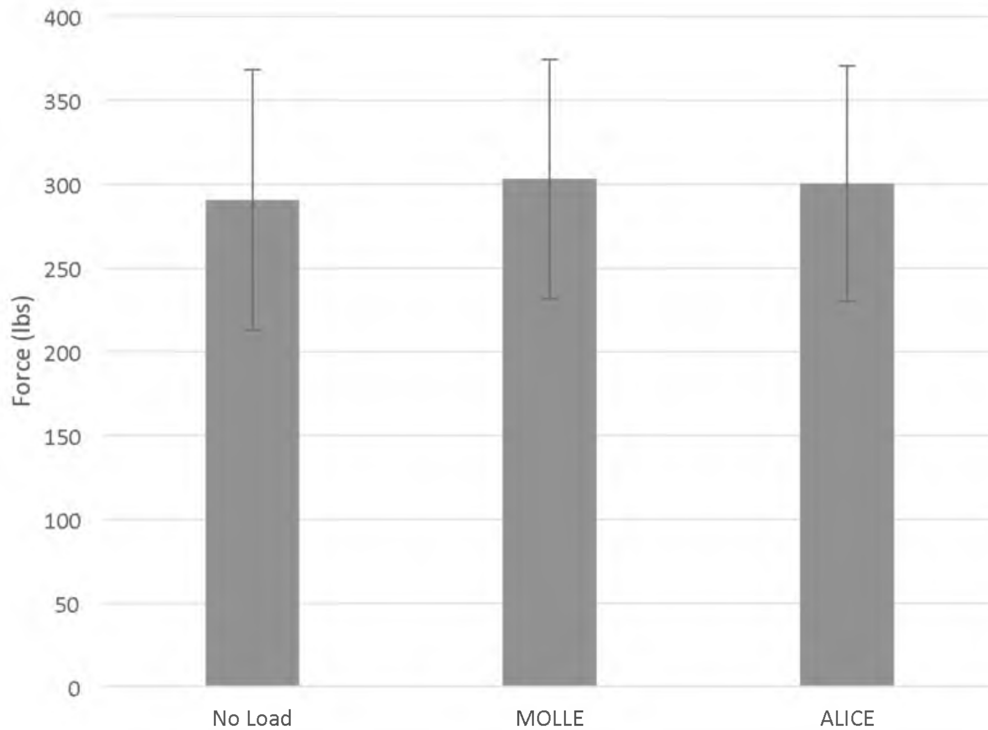

Figure 6-17. Mean Compressive Force on L5/S1 (No Load vs. MOLLE vs. ALICE)

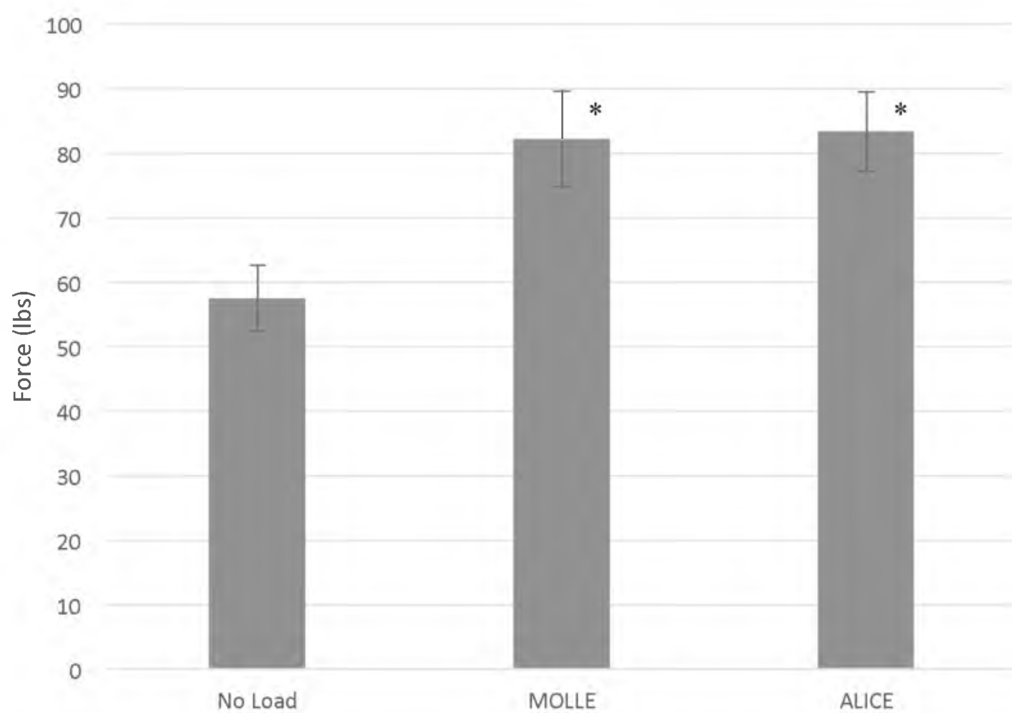

Figure 6-18. Mean Shear Force on L5/S1 (No Load vs. MOLLE vs. ALICE)
* indicates statistical significance (p<.005) compared to No Load

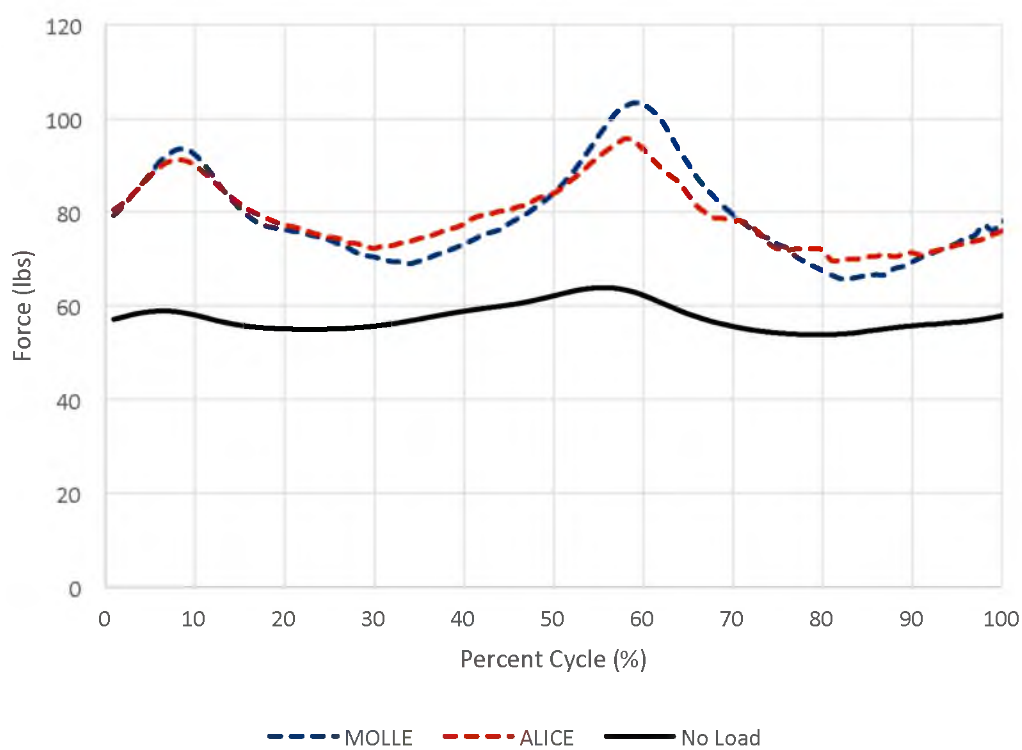

Figure 6-19. Shear Force on L5/S1 Profile (No Load vs. MOLLE vs. ALICE)

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CHAPTER 7

EFFECTS OF PROLONGED EXPOSURE TO CYCLIC LOADING ON L5/S1 LUMBOSACRAL DISC AND POTENTIAL RISK OF DISC FAILURE

The purpose of this chapter was to evaluate the effects of sustained exposure to cyclic loading on the L5/S1 disc during long distance marching with a loaded backpack. A probability model was used to identify the effect, and there was increased probability of disc failure when carrying a backpack on a sandy surface. Besides exposure time to cyclic loading, possible pathways to the spinal discs failure are also discussed.

Introduction

Our result in the previous chapter, showed the compressive force on the L5/S1 disc met NIOSH guidelines. However, this does not necessarily mean that a sustained exposure to cyclic loading is safe. Adams et al. addressed how sustained loading generates pressure concentration on the spine as well as how prolonged stress concentration can induce back pain (1996). Thus, the effects of cyclic loading on the lumbar spine need to be identified to investigate reasons for low back injuries during load carriage.

Callaghan et al. proved disc herniation could be induced with moderate levels of compressive force and low frequency (1 Hz) of cyclic loading. They used porcine spine, which possesses very similar anatomical, geometrical, and functional characteristics with human spine. In their cyclic test result, the likelihood of herniation development was increased with each increment of compressive load magnitude. They suggested "disc herniation is a cumulative process that can result with modest forces if sufficient flexion/extension cycles are applied" (2001).

"The intervertebral disc is a cushion-like structure that transmits loads and provides flexibility to the spine," and the intervertebral discs in the human spine play a significant role in absorbing the compressive load resulting from upper body weight (Chan et al., 2011).

People experience whole body vibration while walking (Matsumoto et al., 1998). An impulsive shock wave is generated at heel strike phase and is transmitted from the lower extremities through the spine (Ogon et al., 2001).

When there is an unexpected vibratory load on the spine, the back muscles dissipate its energy by contracting. Nevertheless, the specific role and mechanism of energy dissipation from the back muscles is still uncertain (Ogon et al., 2001).

It is well-known that load magnitudes, number of repetitions, duration of activity, and frequency are important risk factors related to musculoskeletal disorders from cyclic activities (Lu et al., 2008).

In this chapter, we will estimate disc failure probability when an extended period of cyclic loading is applied on the L5/S1 disc during load carriage.

Method

Probability Model

Schmidt et al. proposed a spine tolerance estimation model based on a Weibull distribution. They insisted spine failure probability from repetitive compressive loading can be estimated using the following equation (2012):

$$\Pi = 1 - \exp\left[-\left(\frac{c}{a_1}\right)^{b_1}\right]$$

Π: disc failure probability

b_1: constant shape parameter (0.438: Male / 0.629: Female)

c: number of cycles

$a_1 = \exp\left(K_0 + K_1 \times Age + K_2 \times Stress\right)$

K_0: constant (19.882: Male / 15.568: Female)

K_1: constant (-0.089: Male / -0.051: Female)

K_2: constant (-4.184: Male / -4.045: Female)

Representative Marching Training Model

To estimate spine tolerance, a 40 km marching distance was assumed, which is one of the most typical marching distances in the Korean army.

Figure 7-1 illustrates the compressive force profiles on the hard surface (solid line) and on the sand surface (dotted line). We found from the profile that each heel strike induces the peak compressive force experienced. The number of cycles (c) is the same with the number of steps. If we assume 4 km/h of marching speed, as army FM (field manual) recommends, trainees would walk 10 h to complete the march.

Results

Probability of Disc Failure on 40 km Marching Training

Average stride length in our research was 0.770± 0.049 m (no load), 0.755±0.048 m (MOLLE), and 0.757±0.048 m (ALICE), respectively. Table 7-1 summarizes the probability of disc failure by loading conditions. The shorter stride length that results when carrying a loaded backpack creates an increased number of cycles in the given distance of marching. The accompanying higher compressive stresses experienced in the low back increase the probability of disc failure. Endplate area of the L5/S1 disc was set as 14 cm^2 based on the research of Schmidt et al. (2012) and then utilized for the disc stress calculation. As a result, backpack carriage increased the possibility of L5/S1 disc failure. MOLLE showed the highest failure probability.

Operational terrain was also an important factor. Thus, the failure probability was also calculated on both the hard and sand surfaces as seen in Table 7-2. Increased peak compressive stresses and the number of cycles resulted in a dramatic increase of the disc failure probability on the sand surface.

Discussion

Possible Pathways to Intervertebral Disc Failure

In the previous chapter, we found the magnitude of the back compressive force met the NIOSH guidelines even when forces were at a maximum due to backpack carriage and surface type. We showed that prolonged exposure to repetitive loading on the L5/S1 create increased chance of the disc failure probability (i.e., walking on the sand surface has 72.9 % higher chance of disc failure in the long term). Figure 7-2 briefly shows some pathways to the pain and injuries based on previous literature.

The human spine undergoes vibration when considering biomechanical aspects of marching and its effects on the lumbar spine. When vibration or cyclic load is applied to the spine, there is disc compression and back muscle fatigue (Pope et al., 1998).

First of all, disc failure can occur from damage accumulation in the annulus fibrosus. Stokes et al. already summarized the possibility of disc failure due to insufficient recovery time from cyclic loading. Specifically, collagen fibers in the annulus fibrosus exhibit an elastic and plastic response due to the vibratory stimuli. The plastic response from the broken covalent bonds of the collagen fibers can induce microdamage to the annulus fibrosus if appropriate recovery time is not allowed. Damage accumulation results in disc failure (2004).

Second, disc injuries and pain can originate in the nucleus pulposus of the disc. Kraemer proved that disc pressure and the disc fluid flow is directly related. There is a disc fluid outflow when a person is standing, sitting or carrying a load, whereas an influx into the disc when one is lying down (1985). Thus, it may imply that the compressive loading on the disc produces an increase in the disc pressure, which would result in the fluid outflow. Disc dehydration after an extended time of exposure will increase proteoglycan concentration and osmotic pressure, so that the disc can absorb more water from external sources. They also noted that the water deficiency may cause disc degeneration and low back pain (1992).

Kraemer observed reversal of the flow around 70-80kpa of intradiscal pressure experimentally (i.e., disc fluid influx below the criteria, outflow above the criteria; 1985). What this means in practical terms is that high compressive forces create an environment where disc fluid outflow is greater than disc fluid inflow, resulting in disc dehydration. So, although there was fluctuation of the compressive loading on the L5/S1 disc along the gait cycle, we may infer that there is a decreased chance of fluid influx on the L5/S1 disc with 28 kg of marching load. In this regard, we assume the fact again that prolonged marching with a heavy backpack can induce low back pain and injuries.

Back pain does not always come from the intervertebral disc. Muscle fatigue can be a main reason for the pain. When the lumbar spine is exposed to vibration, it becomes less stable and shows more deformation under a given load (Liebenson et al., 2009). Thus, a more active “bucking process” (Stokes et al., 2004) will be involved under vibration due to increased range of motion of the spine. It will also alter the back muscle activity (Shin et al., 2010). The altered activity will accelerate muscle fatigue and the muscles will be more vulnerable to improper load distribution and absorption (Pope et al., 1998). It might be another pathway to lower back pain.

The research of Pope et al. also showed back muscle response against unexpected load. The muscles generally overreact against unexpected load for quick stabilization when a sudden load is applied, and show larger latency and greater response amplitude after vibration exposure. They noted that the muscles cannot protect the spine from adverse load, and their forces are added to those of the stimulus due to the latency at many frequencies (1998). Prolonged marching with heavy backpacks might cause similar effects on the back muscles.

More possible pathways exist other than those listed here, and the disc failure mechanism is still unclear. Additional clinical and biomechanical research is required to validate the pathways to disc failure in the future.

Conclusion

Backpack carriage increased the possibility of L5/S1 disc failure from cyclic loading during gait. MOLLE showed a higher failure probability than the other conditions (i.e., no load, ALICE), but additional research is required to see the practical significance. Increased peak compressive stresses and the number of cycles resulted in a dramatic increase of the disc failure probability on the sand surface. Additional research is required to generalize the effect of prolonged exposure to cyclic loading.

Disc failure can occur from damage accumulation in the annulus fibrosus. Disc injuries and back pain can originate from the dehydration of the disc or back muscle fatigue. However, additional clinical and biomechanical research is also required to validate the pathways to disc failure from load carriage and gait.

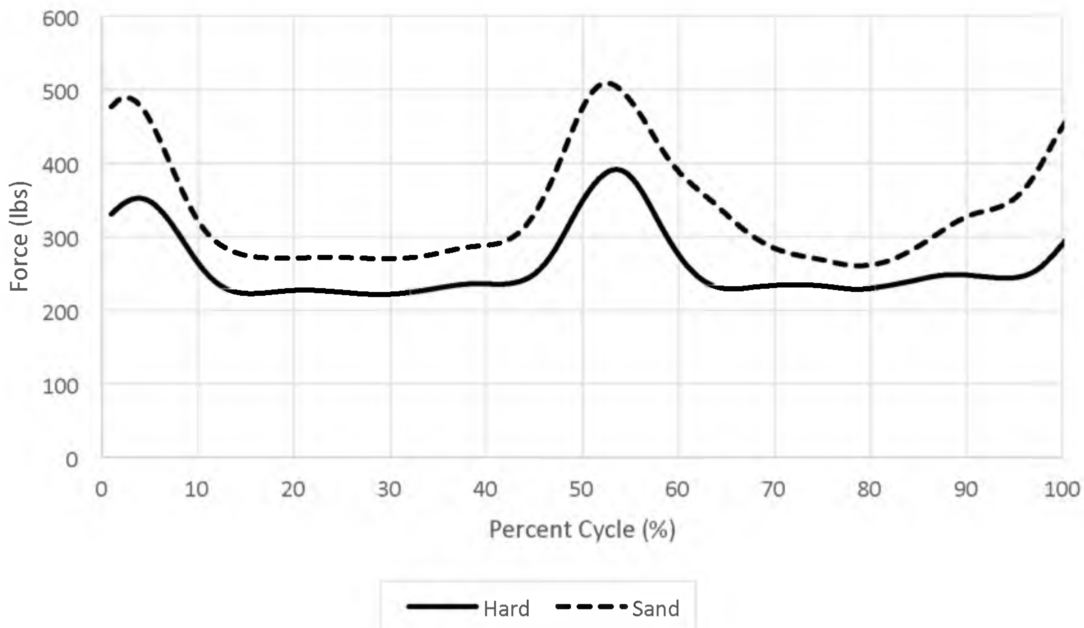

Figure 7-1. Back Compressive Force Profile (Hard vs. Sand)

Table 7-1. Estimation of Disc Failure Probability by Loading Condition

	No Load	MOLLE	ALICE
Stride Length (m)	0.77	0.755	0.757
# of Cycles	51948	52980	52840
Peak Force (N)	1638	1665	1650
Peak Stress (Mpa)	1.17	1.189	1.179
Failure Probability (%)	35.23	36.49	35.89

Table 7-2. Estimation of Disc Failure Probability by Surface Type

	Hard	Sand
Stride Length (m)	0.777	0.745
# of Cycles	51480	53691
Peak Force (N)	1644	2215
Peak Stress (Mpa)	1.17	1.58
Failure Probability (%)	35.12	60.72

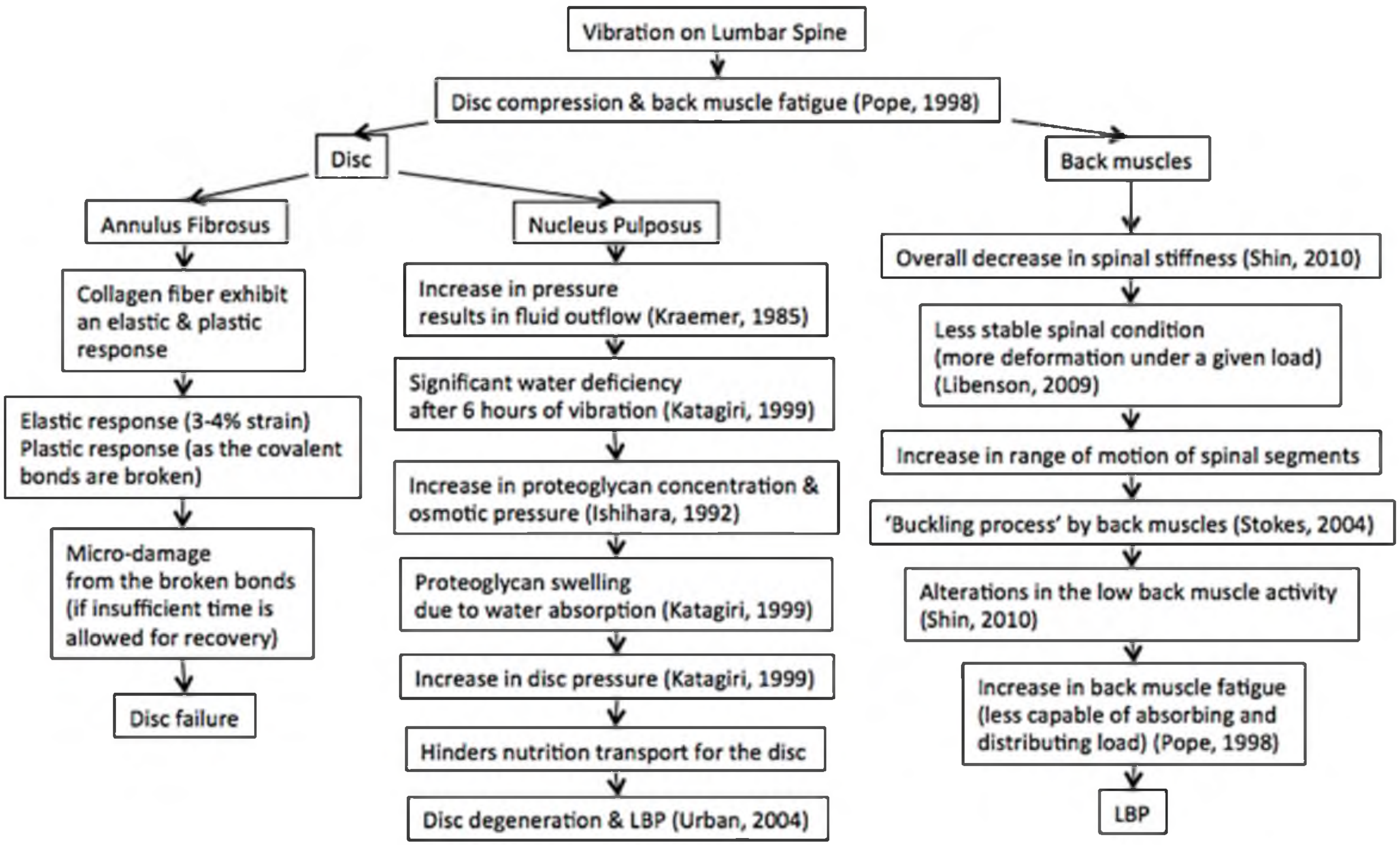

Figure 7-2. Possible Pathways to Intervertebral Disc Failure

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CHAPTER 8

CONCLUSIONS

Summary

Effects of a Backpack Carriage on Soldiers' Body and Potential Injury Risks

Upper body segments were exposed to more deviations, such as greater thorax flexion, greater thorax lateral flexion, and more pelvic anterior tilt, when carrying a backpack than they were under normal walking conditions. These changes resulted in increased shoulder tensions, which increased compressive and shear forces on the L5/S1 disc.

We found asymmetrical profiles of thorax movement in the coronal plane. The asymmetric profile of thoracic motion was more evident when carrying backpacks. This unbalanced movement of the spine might increase the possibility of back pain or injuries. However, the potential risks remained unclear.

With a load, the right shoulder was identified as being higher than the left shoulder. This might be explained from the higher load tolerance of the right shoulder muscles given the assumption that the right side was the dominant shoulder in our study.

There was a significant decrease in transverse pelvic rotation when carrying a backpack, and this can be explained from decreased stride length due to load carriage.

We could thus conclude that a shorter stride length resulted in decreased amplitude of transverse pelvic rotation. Additional whole body analysis is essential, in the future, to investigate complex mechanisms and characteristics of human body motions during load carriage and gait.

We found that there was a slight decrease in erector spinae muscle force when carrying a backpack. In our research, we only measured erector spinae muscle activations, however, we suspect that antagonistic co-contraction of other upper body muscles for spinal stability may partially explain this decrease.

The backpack load exerted stresses on the shoulders. The shoulder forces, due to a backpack load, resulted in increased back compressive and shear forces on the L5/S1 disc with the combinations of upper body weight and erector spinae muscle force. The shoulder forces induced a higher shear force fluctuation when carrying a backpack. The fluctuation might result in a microfracture of the spinal disc after a period of exposure, but it remained unclear in our study since no effective guideline was available.

The back compressive forces were within NIOSH guidelines. However, these forces were not statistically significantly different and we could not rule out the potential risks of prolonged exposure to backpack loading.

Effects of Operational Terrain on Soldiers' Operability and on Commanders' Operational Consideration

METT-TC (mission, enemy, terrain, troops, time, and civilian) factors are key considerations for commanders when planning an operation. One of the main goals of this study was to identify the effect of the terrain factor using a quantitative method.

There were higher thoracic flexion and pelvic anterior tilt in the sagittal plane when walking on the sand surface.

When manoeuvring with and without a backpack, sand terrain created larger back muscle forces. This result can be used to compare forces when soldiers march with full packs along desert terrain versus urban terrain. There was a 91 % increase in peak muscle force on the sand surface when carrying a backpack load compared to the hard surface. This would imply that the typical training – rest cycle in current guidelines should be modified for desert operations.

A contralateral activation pattern of erector spinae muscles was more distinct on the hard surface compared to the sandy surface. This is because the other back muscles (i.e., para-vertebral muscles; left muscle in the case of left heel strike) were also activated on the sand surface and not on the hard surface. This may be because erector spinae muscles activate more to maintain stability on the sand surface.

Overall, two main risk factors were identified in this study for general walking on a sand surface. One was increased amplitude of overall back muscle force. The other was increased activation frequency of each erector spinae muscle to maintain balance.

There were significant increases in compressive and shear lumbar forces on the sand surface. Increased peak compressive stresses and the number of cycles resulted in a dramatic increase of the disc failure probability on the sand surface.

Potential risks performing operations on a desert area were also identified compared to the operations on urban terrains. Increased tensions on shoulders and erector spinae muscles will add to soldiers' fatigue and injury risks during desert operations. Decreased stability on desert terrain may also limit service members' performance and operability.

In this regard, manoeuvring on desert terrain should be minimized and commanders should consider any alternative transportation units available when it is necessary. If the movement is inevitable, they must secure an adequate rest period for the soldiers in the planning stage. As well, personal loadings should be reduced.

Evaluation of Backpack Design and the Importance of Ergonomic and Biomechanical Considerations in the Field of Weapon Systems R&D

Profiles of shoulder reaction force and low back contact force were similar for both backpacks. However, in terms of force magnitude, the MOLLE resulted in significantly higher forces on the shoulder and lower forces along the low back when compared to the ALICE.

Peak forces were identified around 10 % and 60 % of the gait cycle. When there was a significant downward movement of the center of mass (COM) of the backpack, the shoulder reaction force and the low back contact force were increased (i.e., 0 % to 10 % and 50 % to 60 % of the gait cycles). The overall profiles may be related to relative movement of the backpack within a gait cycle. Additional analysis is necessary in the future to generalize the relationship between the COM displacement and the force fluctuations.

Based on our results, a given load of 68 lbs is too heavy to fall within the pain perception level. Possible suggestions for load carriage might thus include reducing the total load and modifying the load distribution between the shoulder and hip area.

There were some limitations in our study. Components of MOLLE (i.e., the assault pack and side pouches) were removed and the hip belt was modified to prevent ASIS marker occlusion. Additional research with different backpack weights and load distribution is needed to analyze the effects of load distribution.

From the cyclic loading point of view, backpack carriage increased the possibility of L5/S1 disc failure. MOLLE showed a higher failure probability than the other conditions (i.e., no load, ALICE). Additional research is required to generalize the effect of prolonged exposure to cyclic loading.

It is obvious that reducing the total load seems to be the simplest solution. It could be accomplished from the ongoing, so called, Future Combat System Modernization Project in the future with the development of weight support systems, load carrying robots (such as HULC® and SMSS), and lightweight personal equipment.

Proper training of soldiers to use the equipment also could be a solution based on individual's physical capability and gait patterns. Other design modifications such as a flexible frame with ventilated padding are also worth consideration if needed.

When developing the new MOLLE, the Korean army did research about the latest trend in backpack designs. Increased operational usability and selected soldiers' subjective preferences were the main priorities used to develop the new backpack. This was done like other personal equipment development processes and failed to include the ergonomic considerations of the soldier throughout the whole processes of development. The US military also uses a similar MOLLE backpack design, but they reported some problems with the design. Additional ergonomic research and analysis need to be performed in the future. Every officer in the field of weapon systems R&D must be aware of the importance of ergonomic considerations from the very beginning of each project.

Limitations

1. All these results were reported based on statistical significance. However, we need to be cautious that the statistical significance does not necessarily mean practical significance in many cases.
2. All participants were allowed to adjust the shoulder straps to provide the best fit and the tension was fixed throughout all trials. However, it was inevitable to modify the MOLLE's hip belt because there were ASIS marker occlusions due to the original belt design. The current study did not quantitatively control the waist belt to minimize the waist belt effect.
3. We assumed that the center of mass of each backpack was located in the geographical center of the backpack, and other modular components of MOLLE, such as side pouches and assault pack, were removed in this study. However, additional research is required to identify the effect of modular components and how changing the location is effects solider stress.

Future Work

1. Our results showed that the differences between two backpacks were not very different from each other. However, we might able to find some practical differences under different weights and modified distributions in the future. Especially, it is important to identify the effect of different load distribution on the shoulder and the hips. Load distribution needs to be considered and discussed in

future research studies.

2. Subjective research is also important when analysing backpack designs. Additional research and analysis is required to identify the relationship between personal preferences and bodily pain/discomfort.

3. Participant selection criteria were set based on military recruitment criteria to reduce variability. Thus, we did not suspect any difference between different ethnic groups in our study because the selection criteria were consistent. However, comparison between each ethnic group might show different results. Additional research is required to evaluate the difference between each group.

4. All participants never used the backpacks, so they represented untrained new recruits in this study. Comparison between the trained and untrained might be important to practically define the marching training procedure. Additional research with different subject selection may be needed in the future.