IMPACT OF BODY ARMOR AND LOAD CARRIAGE

ON LOWER BODY MOVEMENT

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CHAPTER I

INTRODUCTION

Background of Study

Carrying heavy weight is a critical issue for soldiers, both during training and actual duty in a combat zone where agile and speedy performance is a key factor in successful military operations and soldiers' safety. Body armor, which is typically constructed of soft multiple textile layers and heavy rigid ceramic composite plates, and personal carrying loads have been shown to reduce mobility and result in a variety of injuries. For example, according to *Military Technology* (2006), the increased weight and bulkiness of body armor has a noticeable negative impact on agility, comfort and even the soldiers' ability to effectively use their firearms. Birrell, Hooper and Haslam (2007) indicate that carrying loads often reach as heavy as 60% of body weight, which significantly increases injury risk. For instance, an infantry soldier hauls on average between 100 and 150 pounds in addition to the weight of body armor (Leimbach, 2006). Problems resulting from the weight of body armor plus carrying a typical load include foot blisters, knee pain, rucksack palsy, stress fractures, and low back injuries (Knapik, Reynolds, & Harman, 2004). Previous studies also indicate that physiological strain and mobility restriction resulting from wearing body armor and carrying a load can lead to rapid fatigue and dehydration in hot weather conditions (Konitzer, Fargo, Brininger, & Reed, 2008: Manning & Wilson, 2007).

In addition, mobility restriction resulting from wearing body armor and carrying a heavy load can have negative effects on military operations. In particular, mobility of the lower extremities is critically important to an individual soldier's performance and safety since crawling, walking, running and jumping are common basic movements necessary for military operations in combat zones (Man, Swan & Rahmatalla, 2006). Knapik et al. (2004) claim that lightening the weight of loads and improving its distribution can enhance work efficiency by increasing soldiers' mobility and decreasing injury risk.

Mobility has been previously evaluated by assessing subjective perceptions of ease of movement and measurements of two-dimensional range of motion (ROM) and joint angles in standard static postures (Huck, Maganga, & Kim, 1997). However, the measurement of ROM and joint angle in a few standard static postures is limited in scope. Measurements from static postures do not permit evaluation of how carrying a heavy load can change human body movement over time in real working situations, in terms of ergonomic and physiological effects. Thus, a 2-D approach to generating ROM and joint angle measurements is limited since actual human body movements are neither as simple nor as independently identifiable as movement simulated in standard static postures for ROM and joint angle measurement. Rather, the human body moves according to the simultaneous kinematic interaction of each joint, muscle, bone, and force in three-dimensional planes (Watkins, 1984). In this sense, the measurement of ROM and joint angle in static postures cannot adequately capture continuous and interactive changes in body movement over time for a given mobility restriction condition and in measuring the movements required for various working tasks.

Motion capture technology, which tracks human body movement based on a 4-D approach measuring movement of interest in 3-D coordinates over time, is expected to provide more realistic and accurate measurements of changes in body movement resulting from moving while wearing different garment conditions. A motion capture system measures simultaneous, continuous changes in ROM and joint angle at each joint by analyzing a cycle of body movements at each anatomical reference point over time. Motion capture systems have been widely used for biomedical disciplines for clinical and rehabilitation purposes because they can provide numerical data identifying changes in body movement. Gait analysis, a systematic measurement of human locomotion, has used motion capture systems and electromyography (EMG), which is a measurement tool of muscle activation, to diagnose and rehabilitate orthopedic problems in the lower extremity by identifying abnormal working patterns of joints and muscles (Davis, Ounpuu, Tyburski & Gage, 1991). Measurement of plantar pressure and contact area (the plantar surface of the foot that is in contact with the ground) has also been included in gait analysis to provide diagnostic and clinical implications for foot movement-related problems such as skin breakdown and ulcers.

In this vein, measurements of lower body movement through gait analysis based on motion capture, EMG and foot pressure sensor technologies are expected to provide a more comprehensive explanation of the changes in mobility caused by carrying additional weight as well as weight distribution.

Purpose

The overall purpose of this study was to identify impacts that personal body armor and carrying a load have on lower body movement by using a biomechanical approach that relies on motion capture, EMG and foot pressure sensor technologies. Considering that walking is a fundamental and frequently repeated lower body movement, gait analysis was used to identify biomechanical changes of the lower limbs and changes in walking patterns while wearing an outer tactical vest (OTV) and carrying a load on the upper body. A human subject test was conducted to measure quantitative parameters characterizing lower body movements such as distance and temporal parameters of walking, ROM, joint angle, EMG, plantar pressure and the foot's contact area which have been widely investigated in gait analysis. Subjects' perceived discomfort and fatigue was also investigated as qualitative data in addition to the multiple quantitative measurements.

Objectives

First, this study identified the impact of the weight of the OTV and additional carrying loads attached to the OTV on lower body movement by comparing walking patterns, ROMs, joint angle, EMG signals on leg muscles, plantar pressure and contact area of the foot. Second, this study identified the impact of weight distribution for selected weight distributions conditions. Third, this study investigated subjects' perceptions about ease of movement, discomfort and fatigue during walking while wearing different garment treatments with varying weight and weight distributions.

Hypotheses

$H_0$1: There are no significant differences in temporal and distance parameters of walking among subjects wearing treatment garments with varying weight and weight distributions.

$H_0$2: There are no significant differences in ROM for lower body movement among subjects wearing treatment garments with varying weight and weight distributions.

$H_0$3: There are no significant differences in joint angles among subjects wearing treatment garments with varying weight and weight distributions.

$H_0 4$: There are no significant differences in EMG among subjects wearing treatment garments with varying weight and weight distributions.

$H_0 5$: There are no significant differences in plantar pressure among treatment garments with varying weight and weight distributions.

$H_0 6$: There are no significant differences in contact area of the foot among treatment garments with varying weight and weight distributions.

Significance of the Study

This study explored the impact of weight of body armor and carrying load plus weight distribution on lower body movement by investigating multiple variables such as previously specified.

Therefore, first, this study provided a more comprehensive explanation of the changes in mobility of the lower limbs resulting from wearing garment with varying weight and weight distribution conditions as compared to previous studies which have focused on measuring 2D range of motion and joint angle in static postures or subjective perception on ease of movement. Second, the biomechanical approach based on motion capture system, EMG and foot pressure technology and gait analysis in this study may suggest a more accurate and practical methodology for assessing mobility, which may expand the use of the current approach toward mobility evaluation in the field of clothing science. Third, the biomechanical approach used in this study may provide diagnostic implications to minimize possible musculoskeletal injuries and the decrease in mobility caused by weight-bearing, inevitable working conditions in the military. Fourth, the results of this study may provide practical implications for body armor design as well as other types of heavy protective clothing.

Limitations

1. Only one type of ballistic vest, the OTV, was used for this study.

2. Although a few testing methods are available for measuring lower body movement, the Davis protocol gait analysis, one method for performing gait analysis based on passive optical motion capture technology, was used as a test protocol. The passive optical motion capture technology has been widely used in sports science and gait analysis because it does not require cables, which may cause unnatural movement.

3. Load carriage was limited to two weight levels: 20 lbs and 40 lbs and five locations on the OTV. Composition of the load was limited by the use of dumbbells and coins.

4. Sample recruiting was limited to right-handed volunteer ROTC (Reserve Officers' Training Corps) students attending Oklahoma State University who had experience in wearing a ballistic vest. Volunteers with specific physical conditions (height: 5.9 – 6.3 feet, weight: 155 – 230 lbs) were recruited for the human subject test.

Definitions of Terms

Electromyography (EMG): technique for measuring the activation signal produced by skeletal muscles (Wang, Stefano & Allen, 2006).

EMG signal: “a biomedical signal that measures electrical currents generated in muscles during its contraction representing neuromuscular activities” (Reaz, Hussain, & Mohd-Yasin, 2006, p. 11).

Gait analysis: “systematic measurement, description and assessment of those quantities thought to characterize human locomotion” (Davis, Ounpuu, Tyburski, & Gage, 1991, p. 575).

Gait cycle: "the period of time from the point of *initial contact* (also referred to as heel strike) of the subject's foot with the ground to the next point of initial contact for that same limb". A cycle of gait is divided in stance phase and swing phase (Ounpuu, 1994, p. 3).

Joint angle: "the angle between the longitudinal axes of two adjacent body segments" (The Oxford Dictionary of Sports Science & Medicine. Oxford University Press, 2007. Oxford Reference Online.).

Step length: "The distance from a point of contact with the ground of one foot to the following occurrence of the same point of contact with the other foot. The right step length is the distance from the left heel to the right heel when both feet are in contact with the ground." (Ounpuu, 1994, p. 6).

Stride length: "the distance from initial contact of one foot to the following initial contact of the same foot." (Ounpuu, 1994, p. 7). "Stride length can be measured as the length between the heels from one heel strike to the next heel strike on the same side. Two step lengths (left plus right) make one stride length." (Vaughan, B. L. Davis & J. C. O'Connor, 1996, p. 12).

Mobility: "the ease with which an articulation, or a series of articulations, is allowed to move before being restricted by the surrounding structures" (Kreighbaum & Barthels, 1996, p. 64).

Range of motion (ROM): "the total amount of angular displacement through which two adjacent segments may move" (Kreighbaum & Barthels, 1996, p. 64).

Plantar pressure: the force measured per unit area on the force plate or sensor mat when the plantar surface of the foot contacts the force plate or sensor mat (Orlin & McPoil, 2000).

Maximum voluntary isometric contraction test: a standardized method for measurement for maximum muscle strength by forcing a subject's muscle to apply the maximum resistance force against the given matching force so the muscle contracts with relatively constant length and force while no joint movement occurs (Enoka, 2002; Kendall, McCreary, Provance, Rodgers, & Romani, 1993).

CHAPTER II

REVIEW OF LITERATURE

This chapter covers four major topics: 1) motion capture systems, 2) gait analysis, 3) leg muscle dynamics in human gait, 4) foot pressure measurement and 5) impact of weight and weight distribution on the human body.

In the first section, the working mechanism of different motion capture technologies and the benefits of using the system for the measurement of human body movement are introduced. The second section on gait analysis explains how lower body movements while walking are measured and analyzed. Typical parameters characterizing walking patterns such as temporal parameters, distance parameters, joint angle, ROMs and EMG signals are detailed. The third section on leg muscles summarizes the functions and characteristics of major muscles used for walking. In the fourth section, measurement of plantar pressure and contact area, as indicators of foot-related issues are detailed. Finally, the last section reviews previous studies which investigated the effect of weight and weight distribution of carrying loads on lower body movement.

Motion Capture Systems

Motion capture systems are known to simultaneously measure a variety of parameters identifying human body movement such as location, distance, ROM, and velocity in three cardinal planes: frontal, sagittal and transverse (see Figure 1).

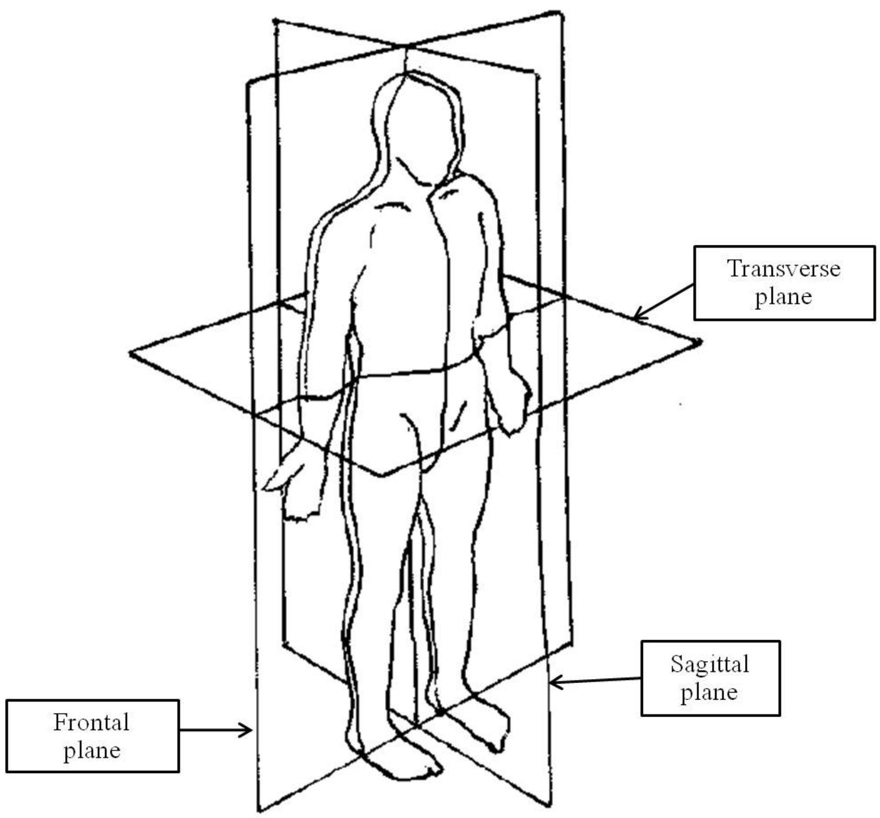


*Figure 1***.** Three cardinal planes. From *Dynamics of Human Gait (2nd ed.)* (p.19), by C. L. Vaughan, B. L. Davis & J. C. O'Connor, 1996, Cape Town, South Africa: Kiboho Publisher. Copyright 1996 by Kiboho Publisher. Adapted with permission.

One of the benefits of using a motion capture system is the accuracy of the resulting data because tracking the changes in location of anatomical points in three planes is more accurate than tracking 2-D measurements in one plane, which may result in the loss of simultaneous movements in the other two planes (Davis, Ounpuu, Tyburski & Deluca, 1991). Currently, different types of motion capture systems including mechanical, electromagnetic, active optical and passive optical motion capture systems are used in different fields depending on desired applications and required working conditions.

A mechanical motion capture system requires subjects to wear a human-shaped straight metal exoskeleton simultaneously moving as the subject moves (Furniss, 2000). A mechanical motion capture system is known for the stable data collection, as compared to other electromagnetic and optical motion capture systems whose data collection can be subject to interference by electromagnetic and light conditions in the testing environment. However, this system is limited to measuring only relative angles between two segments. Therefore, mechanical motion capture systems have limitations in measuring some positions and body orientations (Field, Stirling, Naghdy, & Pan, 2009). Another disadvantage is that the rigidity of the exoskeleton worn over the human body restricts natural body movement (Field et al., 2009).

An electromagnetic motion capture system requires subjects to wear an array of magnetic receivers, which tracks the changes in location of anatomic points. This system is known to be less expensive than optical systems and convenient to use. However, the system is easily subject to interference by environmental magnetic fields and there is a high possibility of incorrect data resulting from magnetic distortion during the test (Furniss, 2000).

Active and passive optical motion capture systems use markers attached to the subject's body and cameras detecting the change in location of the markers. These optical motion capture systems have been widely used for biomedical applications such as sports injury research and gait analyses because these systems do not require cables which may cause unnatural movements (Furniss, 2000).

A passive optical motion capture system uses retroreflective markers and infrared cameras detecting light reflection from the marker surface, while an active optical motion capture system uses light emitting markers. In a passive optical motion capture system, infrared cameras recognize and record the change in position of retroreflective markers attached to the joints during movement for data collection. Collected data from the markers' locations and the video are sent to a processing computer, which calculates predetermined parameters (e.g.: ROM at joints on three planes and walking velocity, etc.) based on numerical data of the markers' location in three-dimensional coordinates by a pre-programmed data processing protocol.

Gait Analysis

Gait analysis, the systematic assessment of human locomotion of the lower body, has been mainly used for clinical applications to diagnose and rehabilitate patients suffering from orthopedic problems caused by a neuromuscular injury or disorder (Davis, Ounpuu, Tyburski & Gage, 1991).

Gait analysis starts with capturing a gait cycle. During walking, one leg bears the body weight and maintains balance by contacting the ground, while the other leg swings forward, and this pattern alternates to achieve movement (Rodgers, 1988). This repetitious, sequential leg movement enables the body to move forward while maintaining stability (Wang et al., 2006). Gait analysis defines a cycle of walking by capturing each leg's sequential movement from heel strike to toe-off and analyzes characteristics of walking patterns such as ROMs at each joint, step length, step width and velocity. Because of its diagnostic benefits, gait analysis is also used as a predictor of running-inflicted injuries, and it provides practical applications to sports science and even running shoe design (Manson, McKean & Stanish, 2008).

The Davis protocol, one method of performing gait analysis, has been recently used for biomedical applications because it provides a high level of accuracy and reliability based on motion capture systems (Davis, Ounpuu, Tyburski & Gage, 1991). The Davis protocol measures continuous change in ROM and temporal and distance parameters during walking by tracking retro-reflective markers attached to a subject's joints. Therefore, the Davis protocol gait analysis typically requires an optical motion capture system, which records human body movement by calculating the change in location of markers in three dimensional coordinates. Collected body movement data captured by infrared cameras are compared with normative walking patterns. A data pool archived from numerous tests with subjects without orthopedic problems to find characteristics of normal and abnormal walking patterns can be used for comparisons.

The Davis protocol gait analysis starts with an anthropometric measurement of the subjects' lower body, which is used as a basis for subsequent comparisons with normative walking pattern data of average persons with similar physical conditions and age. Markers are attached to anatomical points on the subject's body, and then the subject walks a few steps while the infrared cameras detect lower body movement by capturing changes in position of the markers. Walking is videotaped for closer examination and qualitative analysis on the subject's foot dynamics. For data analysis, body movement data collected from the infrared cameras are sent to a processing computer, which defines a walking cycle from heel contact to toe-off for each leg (see Figure 2), then calculates temporal parameters, distance parameters, joint angle and ROMs during a cycle of walking.

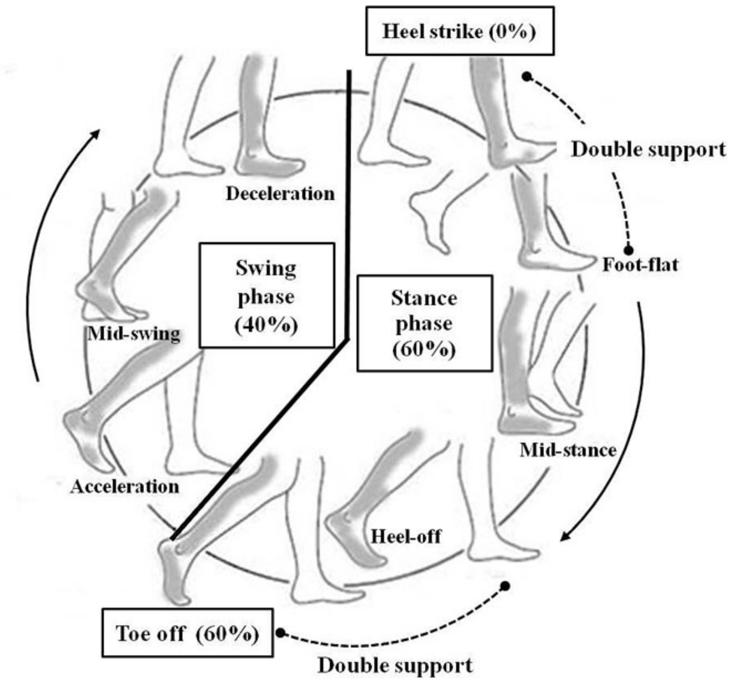


Figure 2. A cycle of walking. From *Dynamics of Human Gait (2nd ed.)* (p.23), by C. L. Vaughan, B. L. Davis & J. C. O'Connor, 1996, Cape Town, South Africa: Kiboho Publisher. Copyright 1996 by Kiboho Publisher. Adapted with permission.

Parameters Characterizing Human Gait

Temporal and Distance Parameters

Temporal parameters include walking velocity, stance phase, swing phase and double support. Walking velocity is walking distance divided by time. Stance phase refers to the period of time when the foot is in contact with the ground, and swing phase indicates the period of time when the foot is not in contact with the ground (Ounpuu, 1994) (see Figure 2). Normative walking is composed of about 60% stance phase and 40% swing phase. Stance phase allows weight-bearing and provides body stability (Rodgers, 1988). As gait speed increases, stance phase decreases and swing phase increases (Mann & Hagy, 1980). On average, stance phase decreases about 30 % for running (from 60% to 30%) and about 40% for sprinting (from 60% to 20%).

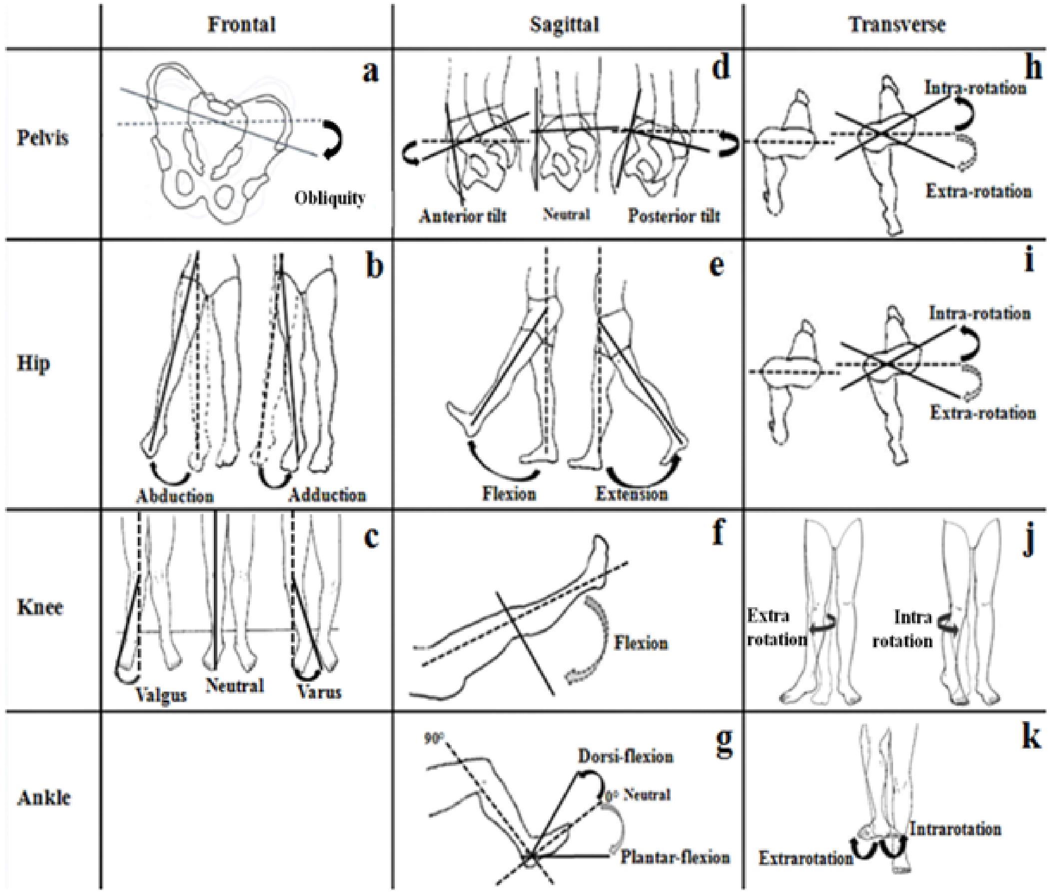
Double support is the period of time when both feet are in contact with the ground (see Figure 2) and decreases with the increase in gait speed (Mann & Hagy, 1980). Therefore, as one walks or runs faster, stance phase and double support typically decrease and swing phase increases, so that the foot contacts the ground for less time. Distance parameters of gait include step length and step width. Step length refers to the distance from one foot's point of contact (heel strike) with the ground to the other foot's point of contact (heel strike) with the ground. Step width indicates the side to side distance between the feet.

Joint Movement: Joint Angle and ROM

In gait analysis, joint angle and ROM for joint movements in the frontal plane have been measured to identify changes in lower limb movements while walking (Davis, Ounpuu, Tyburski & Gage, 1991). Joint angle is defined as "the angle between the longitudinal axes of two adjacent body segments" (The Oxford Dictionary of Sports Science & Medicine. Oxford University Press, 2007. Oxford Reference Online.). ROM is defined as "the total amount of angular displacement through which two adjacent segment may move" by Kreighbaum & Barthels (1996, p. 64). Therefore, ROM can be expressed as a variability in joint angle.

The 3-D approach classifies body movements into three categories: lateral movements, anterior-posterior movements and rotational movements.

Lateral movements are observed in the frontal plane, which vertically splits the body into front and back segments (Vaughan et al., 1996). Major movements in the frontal plane include pelvic obliquity, hip adduction, hip abduction, knee varus and knee valgus. Pelvic obliquity (see Figure 3a) is the angular movement referring to the elevation and depression of the pelvis while one is walking and is observed in the frontal plane.



Legend.

[a] Adapted from *The treatment of gait problems in cerebral palsy* , (p. 105). by J. R. Gage, 2004, London, UK: Mac Keith Press.

[b, j] From *Basic of biomechanics (5th ed.)* , (p. 36 & 38), by S. J. Hall, 2006, New York, NY: McGraw Hill Higher Education. Copyright 2006 by McGraw Hill Higher Education. Adapted with permission.

[c] Adapted from *McGraw-Hill Dictionary of Scientific and Technical Terms.* Retrieved June 18, 2011, from Answers.com Web site: http://www.answers.com/topic/genu-valgum.

[d] Adapted from http://www.pt.ntu.edu.tw/hmchai/Kinesiology/KINspine/PelvicGirdle.htm.

[e] Adapted from http://www.gla.ac.uk/ibls/US/fab/tutorial/anatomy/hip1.html

[f, g] From *Kinesiology: Scientific basis of human motion (9th ed.),* by K. Luttgens & N. Hamilton, 1997, Madison, WI: Brown & Benchmark. Copyright 1997 by McGraw Hill Higher Education. Adapted with permission.

[h, i] From ChiRunning: A revolutionary approach to effortless, injury-free running (p. 100), by K. Dreyer and D. Dreyer, 2009, New York, NY: Simon & Schuster. Copyright 2009 by ChiRunning. Adapted from permission.

[k] Adapted from *Kinesiology: Scientific basis of human motion (7th ed.),* by K. Luttgens & K. Wells, 1982, Philadelphia, PA: CBS College Publishing.

*Figure 3****.*** Characteristic movements in a gait analysis.

Adduction refers to a body segment's movement toward the midline of the body, while abduction indicates the movement of a body segment away from the midline in the plane (see Figure 3b). Knee varus indicates inward angulations of the lower leg below the knee while knee valgus means outward angulations of the lower leg below the knee in the frontal plane (see Figure 3c). Figure 3 shows typical lower body movements observed in the three planes while walking.

Anterior and posterior movements of a body segment are observed in the sagittal plane, which vertically bisects the body into left and right segments (American Academy of Orthopedic Surgeons, 1965). Pelvic tilt (see Figure 3d) refers to the angular rotation of the pelvis in the sagittal plane. Anterior tilt is the upward rotation of the pelvis, and posterior tilt is the downward rotation of the pelvis while walking (American Academy of Orthopedic Surgeons, 1965). In addition to pelvic tilt, five more primary movements occur in the sagittal plane: flexion, extension, hyper-extension, dorsiflexion and plantar flexion (American Academy of Orthopedic Surgeons, 1965). Flexion indicates anterior directional rotation of the head, torso, arm, hand and hip in the sagittal plane and posterior rotation of the legs (see Figure 3e and 3f). Extension is the movement that returns a body part to its anatomical position from the flexion position. Hyperextension, or excessive extension, is the rotational movement beyond anatomical position in the opposite direction to flexion. Hyperextension can take place at the neck and the arm when a joint is overstretched or bent backwards too far, but it does not occur in lower body movement while walking normally. Dorsiflexion is the upward rotation of the foot, and plantarflexion is the downward rotation of the foot at the ankle (see Figure 3g). Pelvic tilt, hip flexion, hip extension, knee flexion and ankle dorsi-plantar flexion are reported to be identifiable joint angles and ROMs in the sagittal plane when describing biomechanical changes in the lower body while walking (Vaughan et al., 1996).

Horizontal rotation of the neck, head, and trunk are observed in the transverse plane, which divides the body horizontally into a top and bottom half (Vaughan et al., 1996). While walking, intra- and extrarotation of the pelvis (see Figure 3h), hip (see Figure 3i), knee (see Figure 3j) and ankle (see Figure 3k) take place. Intra- and extrarotation movement at the knee is called medial- and lateral- rotation. At the ankle, intrarotation is the foot's inward angulations and extrarotation is the foot's outward angulations compared to the walking direction.

In gait analysis, joint angle and ROMs for aforementioned movements in three planes are measured as kinematic parameters during a cycle of walking to identify abnormal gait characteristics.

EMG (Electromyography)

Muscles are moved by electrical signals sent from the brain to the motor unit, which is comprised of motor neurons and bundles of muscle fiber. The electrical signal which activates muscle movement is called action potential and determines various muscle activities such as voluntary contraction and relaxation. When a muscle begins to work, the smaller motor units are first recruited and contracted, then larger motor unites are subsequently recruited. As a muscle works more actively, greater muscle contraction is required. To generate greater muscle contraction, the human body increases both the number of recruited motor units and the firing rate of the motor units (i.e.: frequency of action potential), which significantly increases the amplitude of action potential (Wang et al., 2006).

Electromyography (EMG) detects and records the action potential of a muscle through a pair of polar electrodes attached to the skin on the muscle surface. Collected EMG data have typical parameters such as amplitude, duration, and frequency of action potential; these enable the finding of onset and cessation point of time in the muscle by tracking the activating and discharging moments, which is based on a pre-specified threshold value indicating when the muscle is on and off (Ricamato & Hidler, 2005).

Further, maximum force can be identified and compared by tracking the amplitude over time. For this reason, EMG has been widely used as a diagnostic tool to identify abnormal leg muscle movements by comparing them with normal muscle electrical signals which were collected from subjects without musculoskeletal problems through numerous experiments (Wang, et al., 2006). Wang et al. (2006) showed that each lower extremity muscle has typical characteristic EMG parameters such as onset point, cessation point, and maximum amplitude. Medial gastrocnemius was known to be activated between 9% (onset) and 50% (offset) of a gait cycle, and maximum muscle force was observed at 38% of a gait cycle within a stance phase. On the other hand, tibialis anterior was known to be activated mainly in the swing phase (from 58% to 9% of a gait cycle), and maximum muscle force was found at 95% of a gait cycle (Wang et al., 2006). Van Hedel, Tomatis and Müller (2005) show that the amplitude of lower limb muscle activity increases with an increase in walking speed because the higher walking speed requires greater muscle force to move fast.

Leg Muscle Dynamics in Human Gait

The human leg consists of 28 muscles working as groups for each movement in a gait. The primary role of leg muscles is to generate the accelerating and decelerating forces for safe forward progression (Den Otter, Geurts, Mulder, & Duysens, 2004). Typically, between six and twelve major leg muscles are measured in biomechanics to investigate changes in walking patterns for the given conditions such as walking speed and carrying loads. This is because some muscles located beneath other muscles or bone structures are not measurable, but several large muscles located at the surface can be used as a representative for a muscle group working for each movement in human gait. This is efficient for experimentation and data reduction.

Major Muscles in Lower Body

Upper Leg Muscles

Anterior thigh: A majority of thigh muscles are located on either the tibia or the fibula and work primarily with the knee joint. Thigh muscles function as either the extensor or flexor. Major anterior thigh muscles include sartorius and quadriceps. Sartorius assists knee flexion, hip adduction and hip rotation. Quadriceps consists of rectus femoris (long head and short head), vastus medialis, vastus lateralis, and vastus intermedius. These muscles work together primarily for knee extension. Rectus femoris long head, surrounded by three other vasti muscles, functions additionally as a hip flexor as well as knee extension (Martini, Timmons, & McKindley, 2000) (see Figure 4).

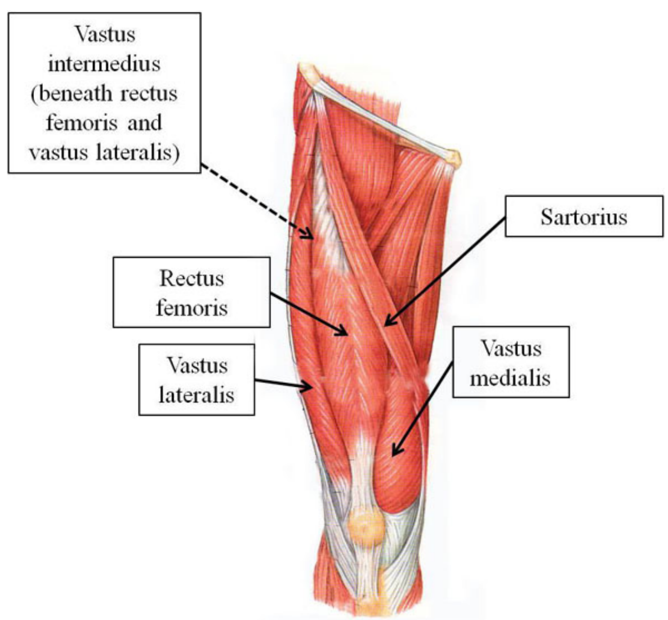


Figure 4. Anterior thigh muscles. Adapted from *Human Anatomy (3rd ed.)*(p. 305) by F. H. Martini, M. J. Timmons & M. P. McKinley, 2000, Upper Saddle River, New Jersey, NY: Prentice Hall.

Table 1 summarizes characteristics and functions of each muscle.

Table 1

Major leg muscles and primary functions

Location	Muscle	Primary function	Remarks
Anterior Thigh	Sartorius	Knee flexion, hip adduction, and hip rotation	Secondary function is cross-legged flexion.
	Rectus femoris	Knee extension	Secondary function: hip flexion
	Vastus medialis		Three vasti group work together for knee flexion as a group. Vastus intermedius is covered by vastus medialis and rectus femoris.
	Vastus lateralis		
	Vastus intermedius		
Posterior thigh	Bicep femoris long head	Knee flexion & hip extension	These three muscles work together as a hamstring group.
	Semitendinosus		
	Semimembranosus		Semimembranosus is located deep beneath skin.
Anterior calf	Tibialis anterior	Dorsiflexion	
Posterior calf	Soleus	Plantarflexion	
	Medial gastrocnemius		
Hip abductor	Gluteus maximus	Hip abduction & stabilization of the pelvis	
	Gluteus medius		
	Gluteus minimus		
	Tensor fascia lata		
Hip adductor	Adductor brevis	Adduction, flexion and medial rotation of the hip	
	Adductor longus		
	Adductor magnus		
	Pectineus		
	Gracilis		

Posterior thigh: Bicep femoris long head, semitendinosus and semimembranosus working together as a group, are known as the hamstrings. These hamstring muscles are known to show a similar EMG pattern during a cycle of walking (see Figure 5). These three muscles primarily control knee flexion and rotation. Furthermore, they assist in hip rotation and hip extension during a cycle of walking (Surface electromyography for the non-invasive assessment of muscles group, 2010).

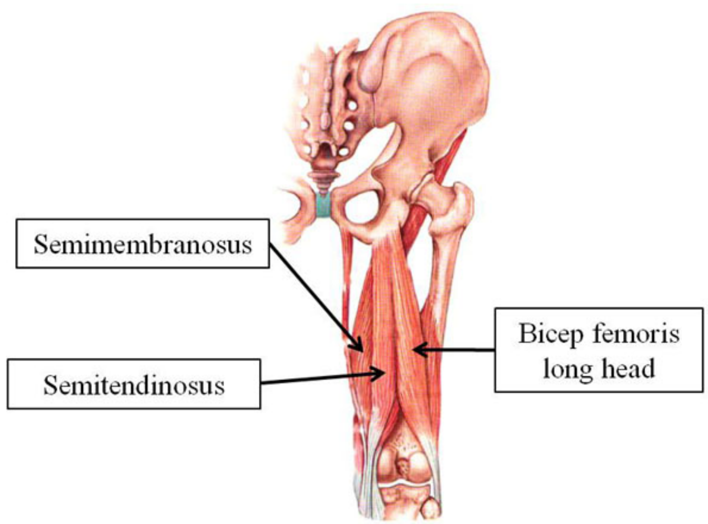


Figure 5. Posterior thigh muscles. Adapted from *Human Anatomy (3rd ed.)*(p. 308) by F. H. Martini, M. J. Timmons & M. P. McKinley, 2000, Upper Saddle River, New Jersey, NY: Prentice Hall.

Anterior and posterior thigh muscles interactively move during walking and running to maximize work efficiency. For instance, when the knee is flexed, the hamstring muscle group functions as a flexor by shortening itself, while rectus femoris is extended as an extensor as shown in Figure 6a. Therefore, a greater amount of tension is applied to rectus femoris, which is often maximized during sprinting. On the other hand, when the leg swings forwards, rectus femoris is shortened and the hamstring is extended in order for the knee to be extended for a wide range of angles (see Figure 6b). Hence, in this motion, a greater amount of tension is applied to the hamstring, which is extremely maximized in hurdling. This is a typical pairwise working pattern of muscles, efficiently using muscle force to significantly decrease the metabolic energy consumption in moving the massive lower limb (Kelley, 1971).

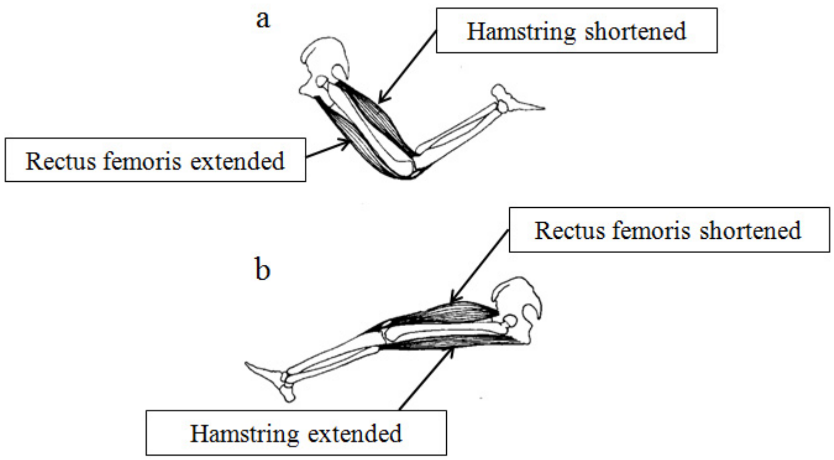


Figure 6. A pairwise interaction between rectus femoris and hamstring in the thigh. Adapted from *Kinesiology: Fundamentals of Motion Description*, (p. 225), by D. L. Kelley, 1971, Englewood Cliffs, New Jersey, NY: Prentice Hall.

Lower Leg Muscles

Between toe-off and subsequent heel-strike, the feet repeat plantarflexion and dorsiflexion to generate sufficient push-off power for necessary leg propulsion toward the next swing phase (Cikajlo & Matjačić, 2007). Lower leg muscles primarily work for dorsiflexion and plantarflexion of the foot during a cycle of walking, although they also control foot inversion and rotation.

Tibialis anterior originates from lateral condyle of tibia and is inserted into medial and plantar surfaces of the first cuneiform and on the base of the first metatarsal. This anterior calf muscle primarily works for dorsiflexion as an extensor. While the leg is not bearing weight in the swing phase, the tibialis anterior dorsal flexes the foot by lifting the middle part of the foot. When the leg touches the ground at the heel strike and begins to bear weight in the stance phase, the tibialis anterior facilitates the foot's mid-stance by pulling the leg towards the foot (Vaughan et al., 1996). In the posterior calf, two major muscles, soleus and gastrocnemius, primarily work for plantarflexion of the foot. Soleus is wider than gastrocnemius, but it is located deeper, thus a majority of the muscle is covered by gastrocnemius. These two muscles unite in a large terminal Achilles tendon attached to the calcaneus (Surface electromyography for the non-invasive assessment of muscles group, 2010). By pulling the Achilles tendon up, these muscles plantarflex the foot.

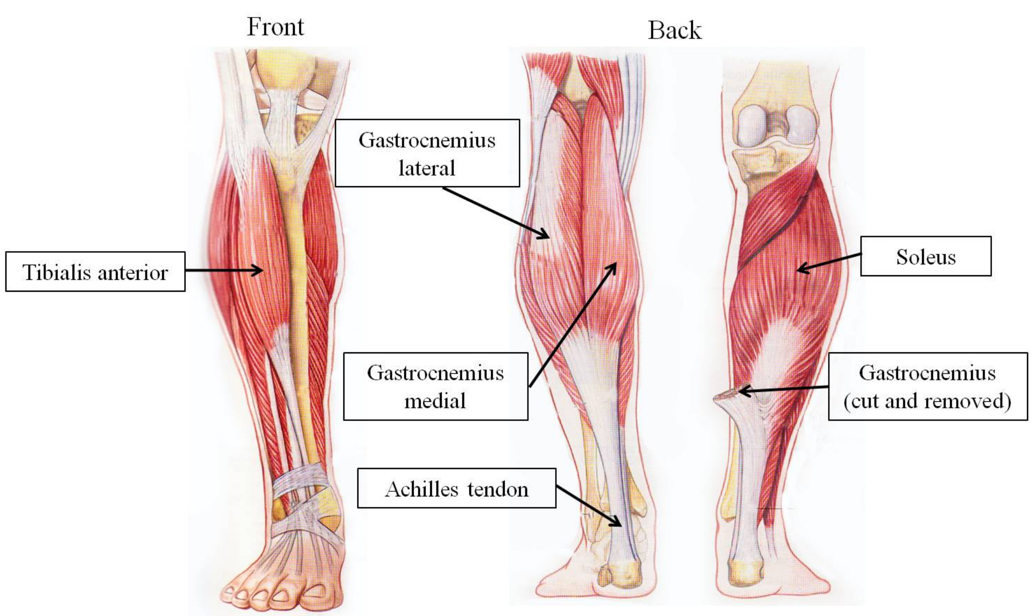


Figure 7. Calf muscles. Adapted from *Human Anatomy (3rd ed.)*(p. 310 & 313) by F. H. Martini, M. J. Timmons & M. P. McKinley, 2000, Upper Saddle River, New Jersey, NY: Prentice Hall.

Since lower leg muscles mainly provide force for propulsion power through dorsi- and plantarflexion, the change in muscle movement in major muscles such as tibialis anterior, soleus and gastrocnemius may influence temporal (e.g.: stance phase, swing phase and double support) and distance parameters (e.g.: step length).

Vaughan et al. (1996) illustrated sequential movements of major leg muscles in a gait cycle (see Figure 8). In Figure 8, the shading indicates the level of muscle activities: black is used to indicate the most active muscle group; stippled is used to indicate an intermediate muscle group; white is used to indicate quiescent (least active) muscle group.

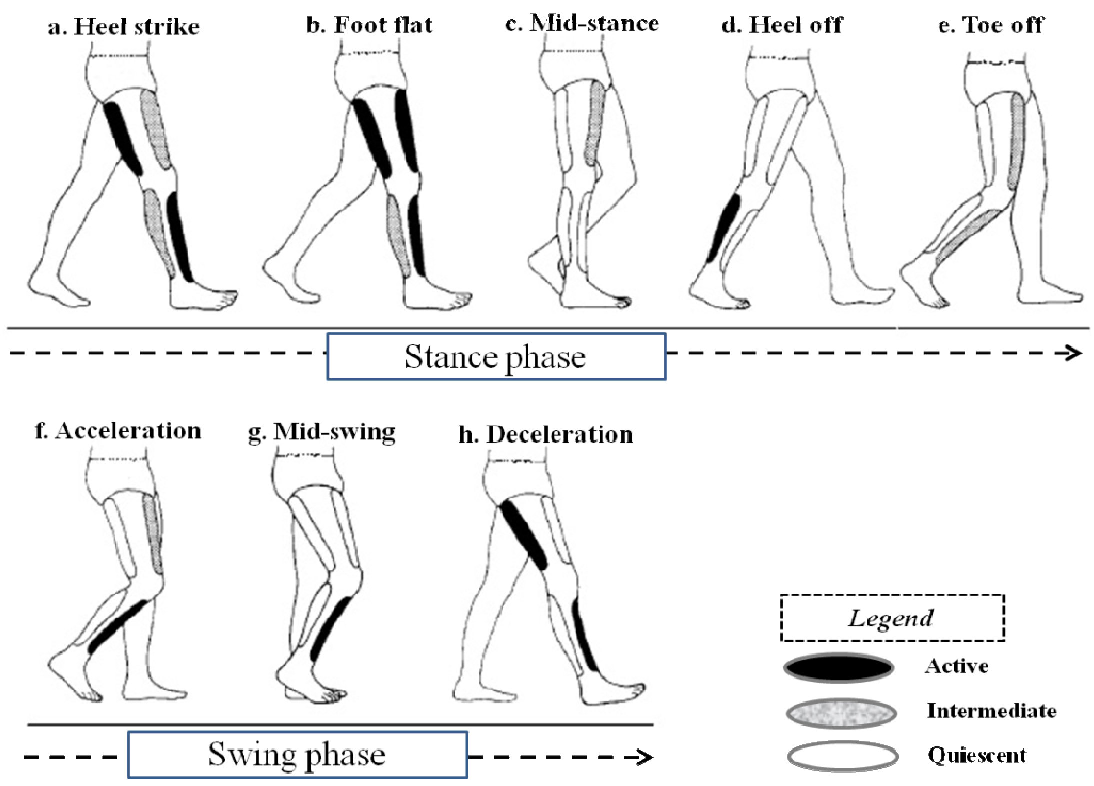
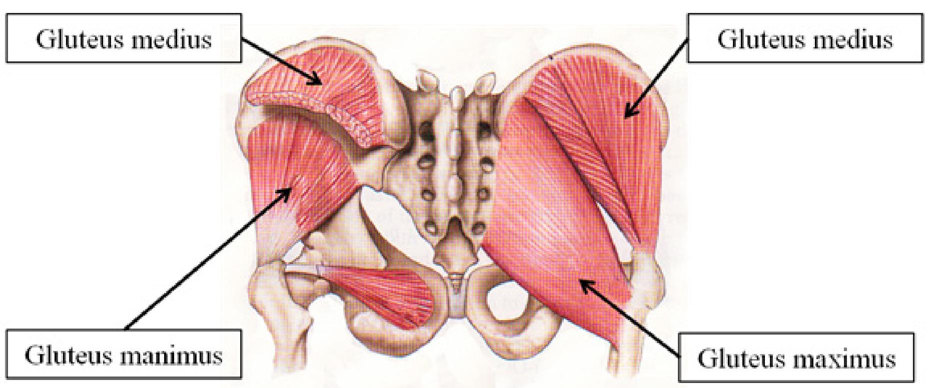


Figure 8. Major leg muscle movements in a gait**.** From *Dynamics of Human Gait (2nd ed.)* (p.54), by C. L. Vaughan, B. L. Davis & J. C. O'Connor, 1996, Cape Town, South Africa: Kiboho Publisher. Copyright 1996 by Kiboho Publisher. Adapted with permission.

Most of the major muscle groups work actively at or around heel strike and toe-off stages (see Figure 8a, 8b, &8h), while major muscle groups are relatively inactive during mid-stance (see Figure 8c) and mid-swing (see Figure 8g). Tibialis anterior actively moves during swing phase for dorsiflexion to keep the toes from dragging on the floor (see Figure 8f, 8g and 8h), while gastrocnemius and soleus work actively for plantarflexion to propel the body forward at the toe-off moment (see Figure 8d).

Hip muscles: Abductors and Adductors

In addition to leg muscles, hip muscles support walking. Major hip muscle groups include hip abductors and adductors. Hip abductors consist of gluteus maximus, gluteus medius, gluteus minimus and tensor fascia lata.



**Note:* tensor fascia lata is not included in this picture

Figure 9. Hip abductors. Adapted from *Human Anatomy (3rd ed.)* (p. 303), by F. H. Martini, M. J. Timmons & M. P. McKinley, 2000, Upper Saddle River, NY: Prentice-Hall.

Abductor group muscles stabilize the pelvis to prevent excessive tilting to unsupported side while in one leg is in a standing position while the other leg swings during walking (Kreighbaum & Barthels, 1996) (see Figure 10). As a result, hip abductor muscles cause the increased compression which increases physical burden to the hip joint and the ligaments surrounding the hip joint (Kreighbaum & Barthels, 1996).

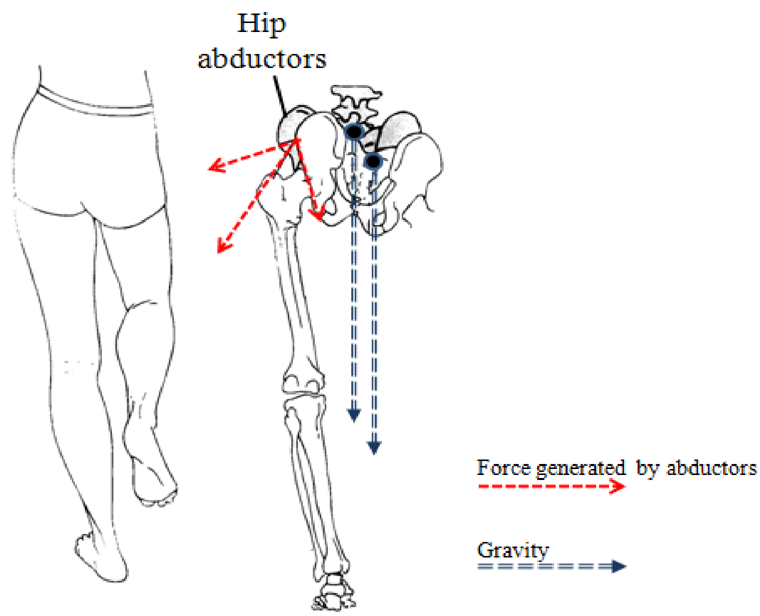


Figure 10. Working mechanism of hip abductors as a stabilizer. Adapted from *Biomechanics: A qualitative approach for studying human movement* (*4th ed*.) (p.195), by E. Kreighbaum & K. M. Barthels, 1996, Needham Heights, MA: A Simon & Schuster Company.

Adductor group consists of adductor brevis, adductor longus, adductor magnus, pectineus and gracilis (see Figure 11). These muscles originate on the pubis and insert on the femur except gracilis, which inserts below the condyle of the tibia (Kreighbaum & Barthels, 1996). Adductor group muscles are responsible for the adduction, flexion and medial rotation of the hip while walking (Kreighbaum & Barthels, 1996).

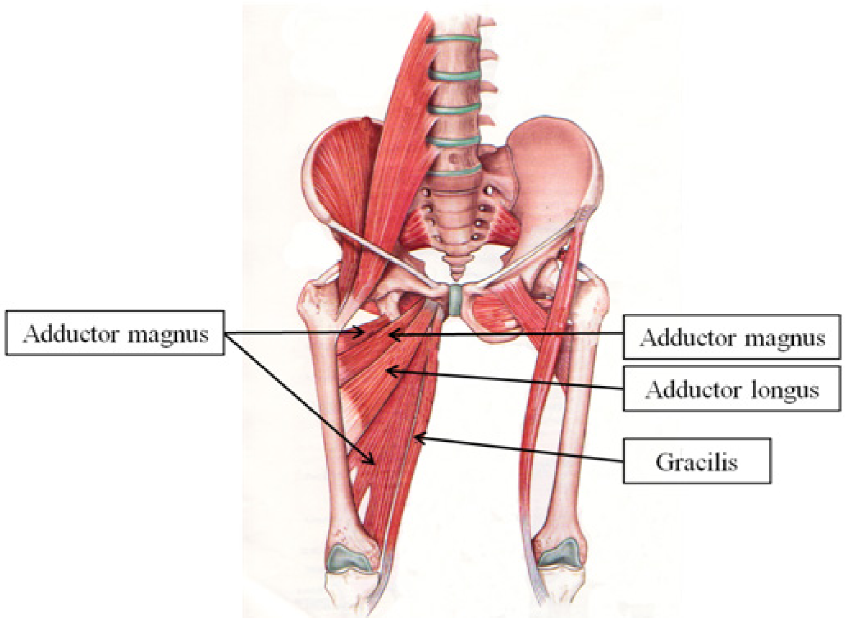


Figure 11. Hip adductors. Adapted from *Human Anatomy* (*3rd ed.)* (p. 304), by F. H. Martini, M. K. Timmons & M. P. McKinley, 2000, Upper Saddle River, NY: Prentice-Hall.

Foot Pressure Measurement: Plantar Pressure and Contact Area

Plantar pressure and contact area have been studied for clinical purposes to diagnose foot movement related problems. Plantar pressure is calculated by dividing force applied to the floor by the area where the foot is in contact with the floor (Zhu, Wertsch, Harris, Loftsgaarden and Price, 1991). In particular, peak plantar pressures, the regional maximum plantar pressures observed in the forefoot, rearfoot and toes, have been extensively studied as an indicator of skin breakdown and ulceration (Zou, Muller, & Lott, 2005). In particular, peak plantar pressure has been investigated to provide clinical implications for people who have diabetes (Caselli, Armstrong, Pham, Veves, and Giurini, 2002; Pitei et al., 1999).

Literature has shown that load carriage and lower body movement have an impact on peak plantar pressure and contact area. Pau, Corona, Leban and Pau (2011), in their backpack study, found that load carriage increased peak plantar pressure and the foot's contact area on the floor, which may cause foot discomfort and foot structure alteration in the case of children who carry backpack for a significantly long period of time. Boulton et al. (1983) showed repeated loading of high peak plantar pressure while walking is a predictor of location of skin breakdown. Orendurff et al. (2008) also showed different peak plantar pressure patterns depending on lower body movements. They demonstrated that cutting and jumping movement in the lower body created greater peak plantar pressure at the heel compared with straight-forward running.

Impact of Weight and Weight Distribution on the Human Body

The impact of weight on body movement is a central issue to improve soldiers' working efficiency and safety, considering that wearing body armor and carrying loads are an inevitable part of military working conditions.

Weight bearing was reported to increase musculoskeletal injury risk and rapid fatigue. Birrell, Hooper and Haslam (2007) show that carrying weight increases the magnitude of impact force at the moment of the heel strike, which is one of the major reasons for overuse injuries such as stress fractures of the tibia and knee joint. Repetitive overloading on bones is known as a direct cause of stress fractures (Knapik et al., 2004). Foot blisters is also one of the common injuries caused by excessive weight resulting in an increased pressure on the skin and more frictions between foot and insole of the boot inside due to higher strain for foot propulsion during walking. Knee pain, low back pain and rucksack palsy are also common orthopedic problems resulting from carrying heavy weight (Knapik et al., 2004). Attwells, Birrell, Hooper and Mansfield (2006) found that carrying heavy load changes standing posture and the head leans more forward to counterbalance loads on the back, which leads to greater muscle strain and tension. Konitzer et al. (2008) claimed that wearing body armor is a direct cause of musculoskeletal pain and negatively impacts combat readiness; the authors pointed to the substantial increase in the reported incidence of musculoskeletal pain with the recent increase in the weight of body armor.

In addition, carrying weight leads to a decrease in mobility by changing body movements. Kinoshita (1985) and Birrell and Haslam (2010) confirmed that an increase in carrying load can significantly decrease walking speed by increasing stance phase, double support, and step width and decreasing swing phase and stride length in order to provide dynamic and static stability. Kinoshita showed that carrying a load weighing more than 20% of body weight significantly changes standing posture and normal walking patterns, which negatively affects walking speed and injury risk. In addition to carrying loads on the trunk, holding a rifle and carrying loads in the hands was also reported to significantly decrease walking speed by restricting natural arm swing, which assists in propelling the body forward (Majumdar, Pal, & Majumdar, 2010).

Carrying weight also significantly affects rapid fatigue by increasing energy expenditure required to complete a task. Teunissenet, Grabowski and Kram (2007) showed that vertical forces to support body weight and horizontal forces to propel body mass are major metabolic costs of walking and running. The weight-bearing conditions require more metabolic energy due to increased vertical and horizontal force necessary in the human body to break the balance of force with gravity in the standing position and move the body forward under the increased gravity. This can induce more rapid fatigue by significantly increasing energy expenditure and adding extra strain to muscles and joints (Smith et al., 2006). Such negative impacts of carrying weight are known to increase in proportion to the increase in weight (Birrell & Haslam, 2010: Kinoshita, 1985).

In addition to the magnitude of weight, weight distribution can also affect mobility. The further weight is located from the “body center of mass” (p. 45), the more energy it costs to carry it (Knapik et al., 2004). For instance, carrying load on the upper body or hip is more efficient than on the hands or feet. Carrying load on the feet costs five to seven times more energy than carrying the same weight load on the upper body (Knapik et al., 2004). Carrying load on the lower leg has a noticeable negative impact on walking speed by limiting leg swing, which has a stronger restriction on walking speed than arm swing, which generates propulsion power (Knapik et al., 2004).

Kinoshita (1985) showed that uniform weight distribution can reduce the negative impact on mobility. In this study, a double supporting backpack with uniform weight distribution to front and back shoulders showed less change in step width, step length and double support, which is closer to normal walking patterns than the equivalent weight backpack placed only on the back side since the weight pulls the body back.

CHAPTER III

METHODOLOGY

This study explored the effect of carrying the weight of body armor as well as the distribution of additional carrying loads on lower body movement using motion capture, EMG and foot pressure sensor technologies. For investigating weight distribution of carrying loads, this study analyzed the difference in lower body movement between the additional weight distributed over multiple areas of the body. A human subject test was conducted to test hypotheses made based on research objectives. Prior to initiating the experiment, the application for conducting a human subject test was approved (Appendix A) by the Oklahoma State University Institutional Review Board for approval.

Independent Variable

The independent variable is a garment condition with seven levels: four levels of weight (1/8 lb, 20 lbs, 40 lbs and 60 lbs) with varying weight distribution. Treatment 1 is a pair of snuggly-fitting 1/8 lb sports shorts (see Figure 12 a). Treatment 2 includes additionally wearing a 20 lb OTV (size Medium) (see Figure 12 b) with front and back ceramic plates (see Figure 12 c). The OTV consists of Cordura ® outer and inner shells, soft armor inserts made of multiple layers of Kevlar, and two ceramic plates. The OTV has webbing on the front and back called a ‘Molle system’, which is designed for load carriage. Equipment such as grenades and walkie-talkies, and pouches containing items required for duty can be attached by using the webbing.

Figure 12. Treatment 1 and 2.

Treatments 3, 4 and 5 have an additional 20 lb carrying load attached to the front of the OTV. In treatment 3 (see Figure 13 a), a 20 lb carrying load was attached to the left vest front while the 20 lb carrying load was attached to the right vest front in treatment 4 (see Figure 13 b). Treatment 5 (see Figure 13 c) has the same 20 lb carrying load, but it is evenly distributed to the left (10 lbs) and right fronts (10 lbs). Figure 13 illustrates treatments 3 through 5.

Figure 13. Treatments 3, 4 and 5.

Treatments 6 and 7 have a 40 lb carrying load attached to the OTV. For treatment 6 (see Figure 14 a), four pouches containing 10 lbs each were attached to the front (left and right) and the back (left and right). In treatment 7 (see Figure 14 b), two pouches containing 20 lbs each were attached to the back (left and right).

front back back

Figure 14. Treatments 6 and 7.

The additional carrying load was manipulated by attaching Molle canteen pouches (see Figure 15) to the webbing of the OTV. For the 10 lb carrying load, a Molle canteen pouch contained two 5 lb dumbbells. For the 20 lb carrying load, a Molle canteen pouch was filled with coins in addition to two 5 lb dumbbells. For the 40 lb carrying load, four Molle canteen pouches containing two 5 lb dumbbells for each were used for treatment 6. Two Molle canteen pouches containing 20 lbs each were used for treatment 7. Each 20 lb Molle canteen pouch was filled with two 5 lb dumbbells and coins.

Figure 15. Molle canteen pouch for load carriage.

Dependent Variables

Dependent variables characterizing lower body movement were measured using motion capture, EMG and foot pressure sensor technologies. The dependent variables are described below.

Temporal and Distance Parameters

Temporal and distance parameters include stance phase (%), swing phase (%), double support (%), stride length (m), walking speed (m/s) and step width (m).

Joint Movement: Joint Angle$_{max}$ and ROM

Each movement at each joint is composed of two different directional motions. For instance, in the case of ankle movement in the sagittal plane, the foot repeats dorsiflexion and plantarflexion at the ankle while walking. If the foot moves upward from the neutral standing position, the movement is dorsiflexion and it is measured as a positive value joint angle. If the foot moves downward from the neutral standing position, this movement is plantarflexion and it is measured as a negative value joint angle. Likewise, all movements at each joint were measured as a form of joint angle with either a positive or a negative value depending on the movement direction. ROM at each joint was defined as “the total amount of angular displacement through which two adjacent segments may move” (Kreighbaum & Barthels, 1996, p. 64) as shown in Figure 16. Therefore, 10 ROMs and 19 joint angle$_{max}$ shown in Table 2 were measured to identify the garment effect on joint movement while walking.

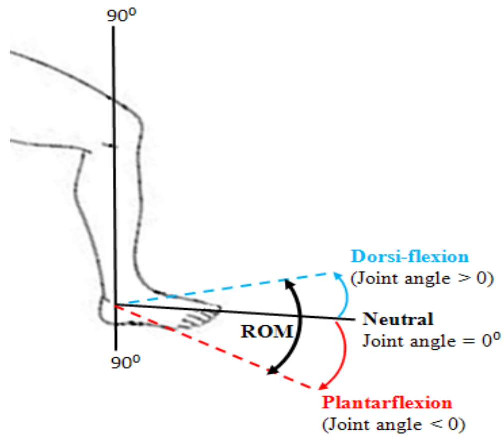


Figure 16. Joint angle and ROM of the foot at the ankle in the sagittal plane. From Kinesiology: Scientific Basis of Human Motion (9th ed.), by K, Luttgens & N. Hamilton, 1997, Madison, WI: Brown & Benchmark. Copyright 1997 by McGraw Hill Higher Education. Adapted with permission.

Table 2

Measurement of ROM and Joint Angles

Joint	Plane	Movement	Directional movement
Pelvis	Frontal	1) Pelvic obliquity	1-1) Upward obliquity (Joint angle > 0) 1-2) Downward obliquity (Joint angle < 0)
	Sagittal	2) Pelvic tilt	2-1) Anterior tilt (Joint angle > 0) 2-2) Posterior tilt (Joint angle < 0)
	Transverse	3) Pelvic rotation	3-1) Intrarotation (Joint angle > 0) 3-2) Extrarotation (Joint angle < 0)
Hip	Frontal	4) Hip adduction & abduction	4-1) Adduction (Joint angle > 0) 4-2) Abduction (Joint angle < 0)
	Sagittal	5) Hip flexion and extension	5-1) Flexion (Joint angle > 0) 5-2) Extension (Joint angle < 0)
	Transverse	6) Hip rotation	6-1) Intrarotation (Joint angle > 0) 6-2) Extrarotation (Joint angle < 0)
Knee	Sagittal	7) Knee flexion	7-1) Flexion (Joint angle > 0)
	Transverse	8) Knee rotation	8-1) Intrarotation (Joint angle > 0) 8-2) Extrarotation (Joint angle < 0)
Ankle	Sagittal	9) Ankle flexion	9-1) Dorsiflexion (Joint angle > 0) 9-2) Plantarflexion (Joint angle < 0)
	Transverse	10) Ankle rotation	10-1) Intrarotation (Joint angle > 0) 10-2) Extrarotation (Joint angle < 0)

EMG

The amplitude (mV) of the EMG signal was collected on four muscles on both right and left legs during the walking cycle: rectus femoris, bicep femoris long head, tibialis anterior and medial gastrocnemius. Considering subjects' individual differences in muscle strength, the amplitude of the EMG signal was normalized by dividing each EMG amplitude measurement by the subject's maximum amplitude of each muscle measured in maximum voluntary isometric contraction (MVIC) test. Therefore, the normalized amplitude was given as a percent of maximum amplitude for each subject, which allowed the researcher to identify changes in each subject regardless of individual differences in muscle strength. To identify the change in amplitude under different garment treatments, the maximum value of normalized EMG was statistically analyzed.

Plantar Pressure and Contact Area

Plantar pressure and contact area were measured in the barefoot condition on a foot pressure sensor mat while wearing different garment treatments. To identify the change in plantar pressure, the peak plantar pressure (PPP) at the forefoot, rear foot and toes and average plantar pressure were recorded in addition to contact area.

Experimental Procedures

Subjects and Sampling

To clearly investigate the significant effect of an independent variable on dependent variables, this study controlled the following physical attributes of the sample group: gender, height, weight, and handedness.

Healthy male ROTC student volunteers with experience in wearing a ballistic vest and with no history of an orthopedic disorder were recruited. Participants were sought between 5.9 and 6.3 feet tall and between 155 and 230 pounds to allow for an appropriate fit of the test garment, a size medium outer tactical vest (OTV). Right handed participants were sought to control possible effect of handedness. Fliers including information about this study (Appendix B) were posted at the Oklahoma State University ROTC office and on campus for recruiting volunteers. In addition, with instructors' permission, in-person contacts were used to recruit subjects using a prepared script (Appendix C) during ROTC classes. The researcher met volunteers at the lobby of the Colvin Center on campus. Volunteers wore a medium size OTV on the bare upper body and the researcher visually assessed the fit of the vest. Visual fit assessment included the following requirements:

1) The OTV must cover the torso area down to the bottom of the rib cage, but must not cover the waistline of the subject, where the markers will be placed during the experiment.

2) The vest must fit the subject by using adjustable straps with Velcro closure without leaving any areas on the side of the body uncovered.

3) The OTV armhole must not interfere with the natural swing of the arm, which occurs if the armhole is too small or too high in the underarm area.

If the volunteer met the fit requirements for the OTV, then the researcher also reviewed the volunteer's prepared medical history check list (Appendix D) to determine if the volunteer's self-reported physical condition was appropriate for the study. If the volunteer had no self-reported orthopedic disorder, the volunteer was considered to be an eligible participant. Based on the aforementioned process, seven right-handed volunteers (age: 21.3 ± 1.1, height: 6 feet ± 1 ⅞ inches, weight: 200 ± 24 lb) were recruited as shown in Figure 17. All recruited volunteers signed the informed consent form (Appendix E).

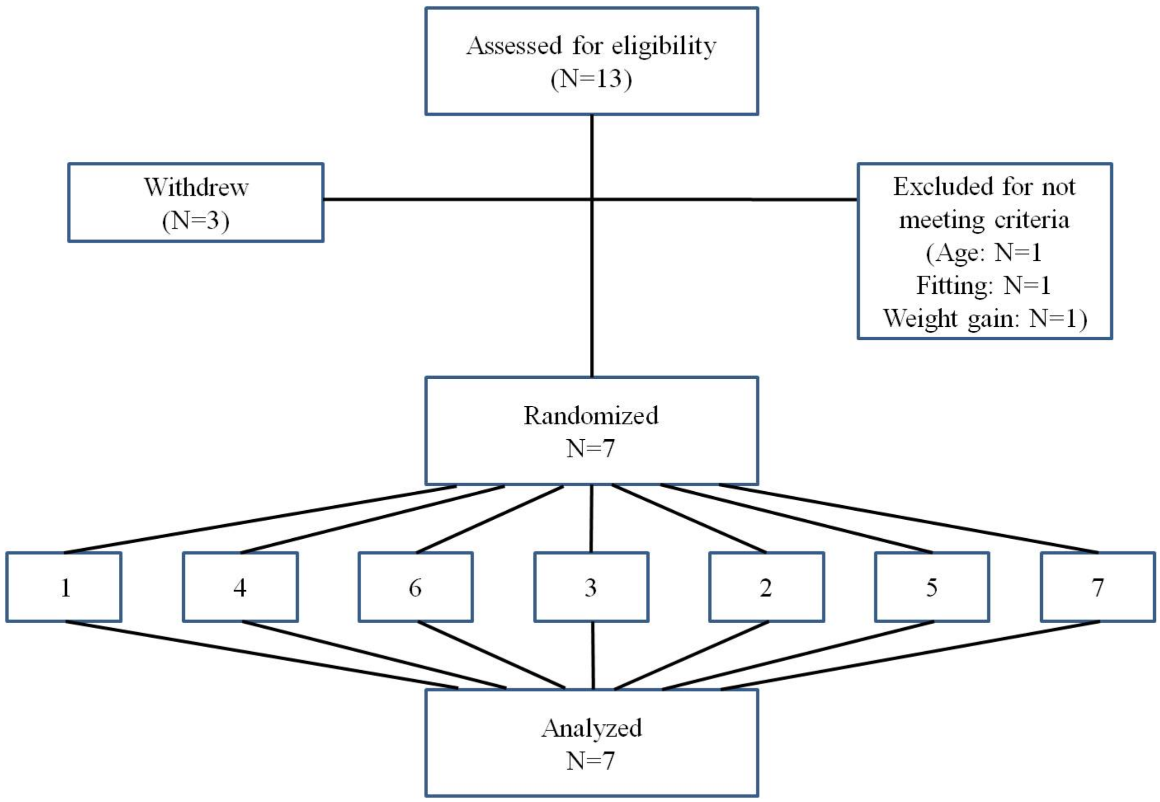


Figure 17. Sampling procedure

Pre-Experiment Protocol

Pre-experiment and experiment procedures were conducted at the IPART (Institute for Protective Apparel Research and Technology) Laboratory at Venture I in the Oklahoma Technology and Research Park located at 1110 S. Innovation Way, Stillwater.

Anthropometric Measurements

Since the experiment was based on the Davis protocol gait analysis, the participants donned tight-fitting shorts for obtaining accurate anthropometric measurements. Then the participant was asked to assume a horizontal position on the medical bed and five anthropometric measurements were taken by using a pelvimeter and tape measure as given in Table 3. Approximately ten minutes were required for this process.

Table 3

Anthropometric measurements

Parameter	Description
Pelvis height	Distance between *anterior superior iliac spine (ASIS) and *greater trochanter line
Pelvis width	Distance between two ASIS
Knee diameter	Distance between *condyles of a knee
Ankle diameter	Distance between *malleolus
Leg length	Distance between ASIS and the mid-point between malleolus

Note. *Anterior superior iliac spine (ASIS): Extremity of the iliac crest of the pelvis, which provides attachment for the inguinal ligament, sartorius muscle, and tensor fasciae latae
*Greater trochanter: The greater trochanter of the femur is a large, irregular, quadrilateral eminence and a part of the skeletal system
* Condyle: The knuckle of any joint, a round projection, rounded articular area
* Malleolus: The bony prominence on each side of the ankle

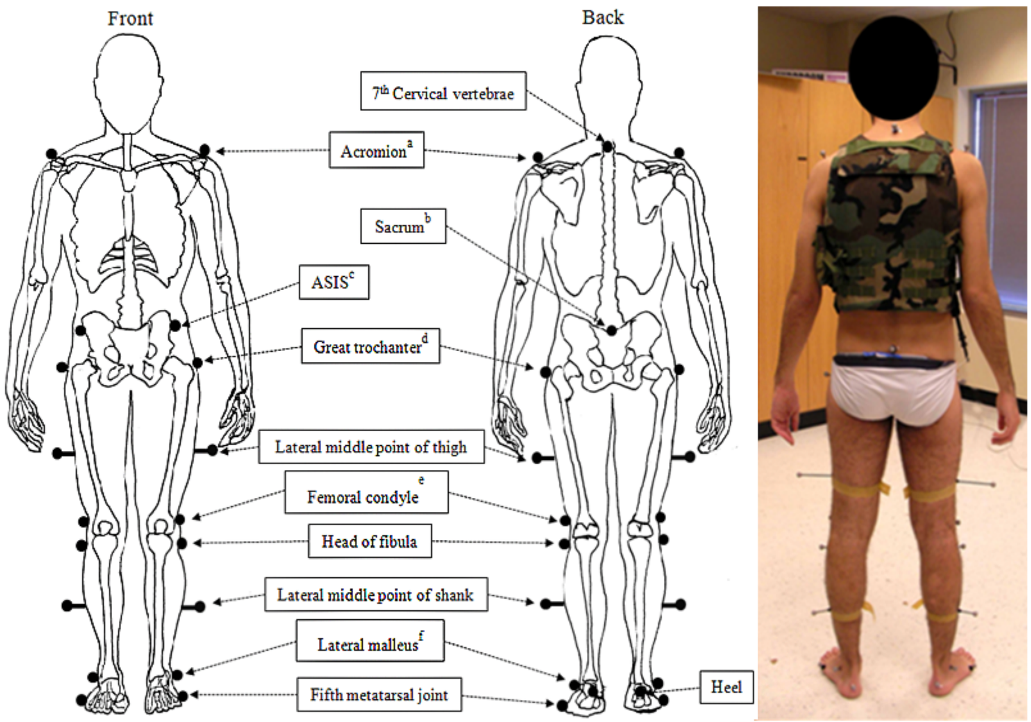
Marker and EMG Placement

To measure and record subject's body movement, the BTS Smart-D Motion Capture System® (BTS Bioengineering, Milano) was used. The BTS Smart-D Motion Capture System® includes retro-reflective markers (10 mm diameter) (see Figure 18), infrared cameras and a processing computer (see Figure 19). Spherical-shaped retroreflective markers with an adhesive surface were attached to each subject's skin at 22 anatomical points on the shoulder and lower body as shown in Figure 20.

Figure 18. Retroreflective markers.

*Figure 19***.** Infrared cameras and a processing computer. Source: Btsbioengineering.com.

For EMG placement, the skin area, where the EMG probes were to be placed, was shaved by the subject with a razor in order to minimize possible signal noise caused by body hair. Then the subject rubbed the shaved skin area with ECG skin preparation gel (Nuprep®, Weaver and Company) to lower skin impedance which may cause incorrect data collection. Then the subject wiped the gel with a soft paper towel and cleaned the shaved skin area with an alcohol pad provided by the researcher. The subject waited for the alcohol to completely evaporate to make sure that the skin area was dry before placing the EMG electrode.



Note: Acromion [a]: The projection of the scapula (the shoulder blade) that forms the point of the shoulder.

Sacrum [b]: The large heavy bone at the base of the spine, which is made up of fused sacral vertebrae and is located in the vertebral column, between the lumbar vertebrae and the coccyx.

ASIS [c]: Anterior superior illiac spine, Extremity of the iliac crest of the pelvis, which provides attachment for the inguinal ligament, sartorius muscle, and tensor fasciae latae.

Great trochanter [d]: A large, irregular, quadrilateral eminence of femur

Femoral condyle [e]: The rounded projected articular area in femor

Lateral malleus [f]: Bony prominence on each side of the ankle

Figure 20. Marker placement.

Eight EMG electrodes, Arbo ™ 124 SG Electrodes (Coviden Commercial Ltd., UK) (see Figure 21) were placed on the skin of four muscles on each leg: Rectus femoris (see Figure 22), bicep femoris long head (see Figure 23), tibialis anterior (see Figure 24), and medial gastrocnemius (see Figure 25) on the left and right leg. Eight wireless EMG probes (see Figure 26) from BTS Bioengineering (Milano, Italy) were snapped onto disposable EMG electrodes (24 mm diameter) to detect the electronic signal of muscle activity. Approximately 20 minutes were taken to place markers and EMG probes (see Figure 26).

Figure 21. Arbo ™ 124 SG Electrode. Source: bi-medical.com.

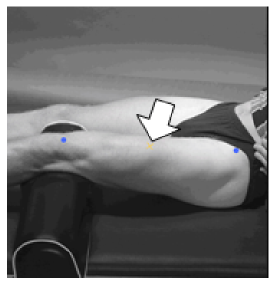


Figure 22. EMG probe placement on rectus femoris. Source: Seniam.com.

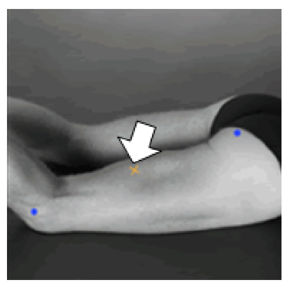


Figure 23. EMG probe placement on bicep femoris long head. Source: Seniam.com.

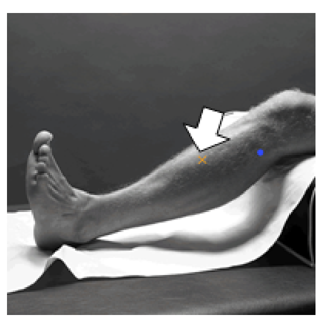


Figure 24. EMG probe placement on tibialis anterior. Source: Seniam.com.

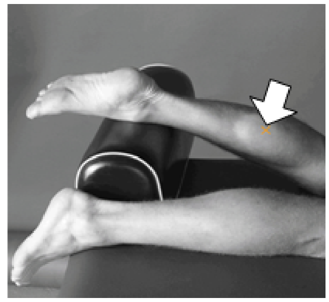


Figure 25. EMG probe placement on medial gastrocnemius. Source: Seniam.com.

Figure 26. Surface EMG probes. Source: Btsbioengineering.com.

Experiment Protocol

Maximum Voluntary Isometric Contraction (MVIC) Test

After placing EMG probes and electrodes on leg muscles to be measured, the MVIC test was conducted to obtain the maximum amplitude of EMG signal for each muscle, which was used as the baseline for EMG normalization. A series of MVIC tests were performed based on the methodology suggested by Kendall, McCreary, Provance, Rodgers, & Romani (1993). EMG signal was measured for each muscle for five seconds in the MVIC condition and the subject took a rest for 30 seconds after completing each MVIC test. The MVIC conditions were created based on Kendall et al. (1993)'s method described as follows.

MVIC test for Rectus femoris: As shown in Figure 27, the subject sat with one knee straight and the other bent over the side of medical bed. The subject's hands grasped the edge of medical bed during the test. The subject leaned backward to relieve hamstring muscle tension. Then the researcher applied force to the straight leg above the ankle in the direction of flexion to bend it. The subject was asked to resist the force by trying to keep the leg straight in the direction of extension, which created the MVIC condition. EMG signal was measured for five seconds in the MVIC condition.

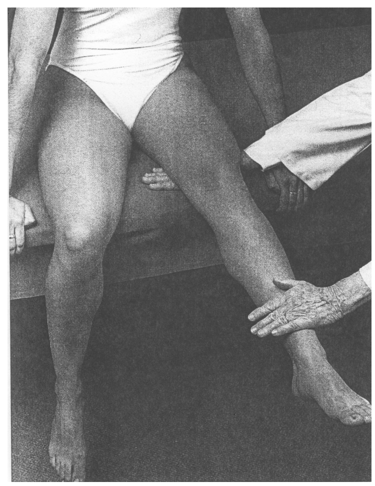


Figure 27. MVIC test for Rectus femoris. From *Muscles: Testing and Function. (4th ed.),* (p. 421), by F. P. Kendall, E. K. McCreary, P. G. Provance, M. M. Rodgers, & W. A. Romani, 1993, Baltimore, MD: Williams & Wilkins.

MVIC test for Bicep femoris long head: As shown in Figure 28, the subject laid prone on the medical bed with one knee flexed between 50º and 70 º, the leg slightly laterally rotated (toes pointing laterally). The other leg, which was not measured, stayed naturally straight on the medical bed. The researcher applied force in the direction of knee extension and the subject was asked to resist the force by trying to flex the knee in order to maintain the leg in the initial posture.

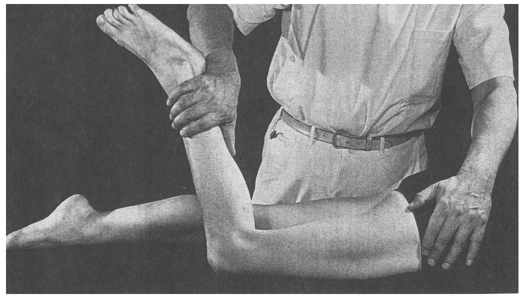


Figure 28. MVIC test for Bicep femoris long head. From *Muscles: Testing and Function. (4th ed.)*, (p. 419), by F. P. Kendall, E. K. McCreary, P. G. Provance, M. M. Rodgers, & W. A. Romani, 1993, Baltimore, MD: Williams & Wilkins.

MVIC test for Tibialis anterior: the subject laid supine and straightened one leg with his foot inversed as shown in Figure 29. The researcher held the leg above the ankle with one hand and applied force in the direction of plantar flexion of the ankle and eversion of the foot with the other hand. The subject resisted the force in the direction of dorsiflexion of the ankle and inversion of the foot.

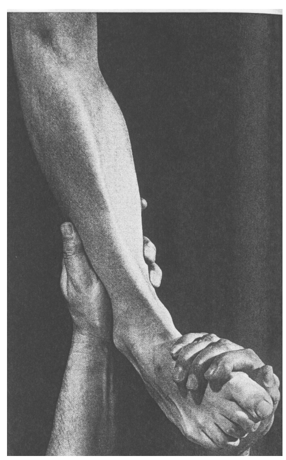


Figure 29. MVIC test for Tibialis anterior. From *Muscles: Testing and Function. (4th ed.)*, (p. 410), by F. P. Kendall, E. K. McCreary, P. G. Provance, M. M. Rodgers, & W. A. Romani, 1993, Baltimore, MD: Williams & Wilkins.

MVIC test for medial gastrocnemius: the subject stood tippy toe on the leg to be measured, while the other leg was not in contact with the floor (see Figure 30). The subject was allowed to use one or two fingers on the edge of the medical bed to maintain postural stability. However, the subject was not allowed to lean his body weight on their finger(s). The knee in the leg to be measured was fully extended. The subject was asked to rise on his toes as high as possible and maintain the same posture for five seconds while the EMG signal was measured.

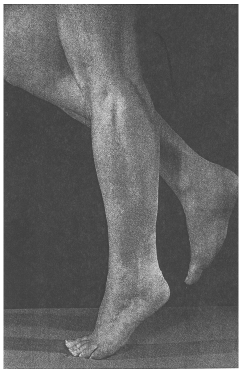


Figure 30. MVIC test for medial gastrocnemius. From *Muscles: Testing and Function. (4th ed.),* (p. 415), by F. P. Kendall, E. K. McCreary, P. G. Provance, M. M. Rodgers, & W. A. Romani, 1993, Baltimore, MD: Williams & Wilkins.

The MVIC test was conducted for eight muscles in the order shown in Figure 31.

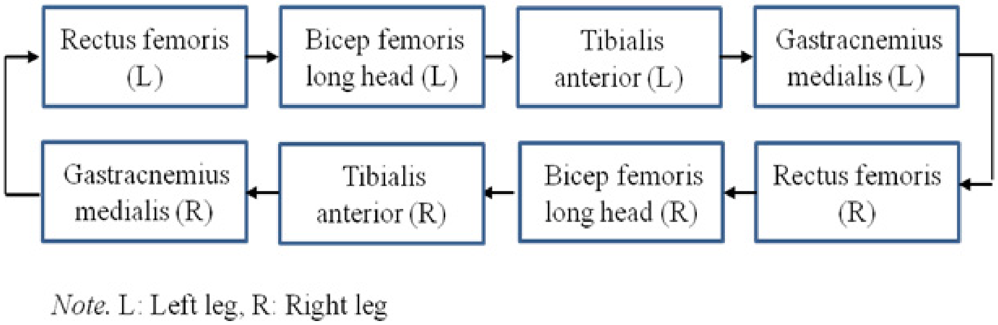


Figure 31. Cycle of MVIC test.

Each MVIC test for each muscle took about 40 seconds including a 30 second rest time before the following test, thus measuring all eight muscles took approximately five minutes. To obtain a reliable measurement of the maximum amplitude for each muscle, the MVIC tests were performed three times, and the average value of the maximum amplitude for three measurements was used for normalization for each muscle. Therefore, a total time of MVIC tests was about fifteen minutes.

Measurement of Plantar Pressure and Contact Area

After the subject completed the MVIC test, the researcher placed a foot pressure mat (34 inch x 15 inch) on the floor. The subject walked four steps on the foot pressure sensor mat (Walkway ™ system, Tekscan Inc., MA, US) (see Figure 32). Each subject performed tests for each garment condition. The order of the garment conditions was set using a Latin Square Design (see Figure 33). Note each garment condition was evaluated once per subject and once per period. The plantar pressure measurement was repeated four times for each garment condition. The walking test for each garment condition took about twelve minutes including the time for system preparation and data saving.

Figure 32. Walkway ™ System. Source: Tekscan.com.

Subject

Period	S1	S2	S3	S4	S5	S6	S7
1	T1	T4	T6	T3	T2	T5	T7
2	T4	T6	T3	T2	T5	T7	T1
3	T6	T3	T2	T5	T7	T1	T4
4	T3	T2	T5	T7	T1	T4	T6
5	T2	T5	T7	T1	T4	T6	T3
6	T5	T7	T1	T4	T6	T3	T2
7	T7	T1	T4	T6	T3	T2	T5

Note. S# identifies the subject; T# identifies garment treatment

Figure 33. Order of walking test based on Latin Square Design.

Walking Test

After the subject completed four repetitions of the walking test on the foot pressure sensor mat, subjects walked about five meters (16 feet) in a line naturally with bare feet in a walking area of about 4 m (13 feet) x 4 m (13 feet) at a self-preferred speed while wearing each garment and carrying load on the upper body (see Figure 34).

Eight infrared cameras recorded the location of markers for data collection while the EMG probes measured the electrical muscle signals during walking. For data analysis, a processing computer defined a cycle of walking based on data captured during four steps. In other words, a cycle of walking consists of four steps.

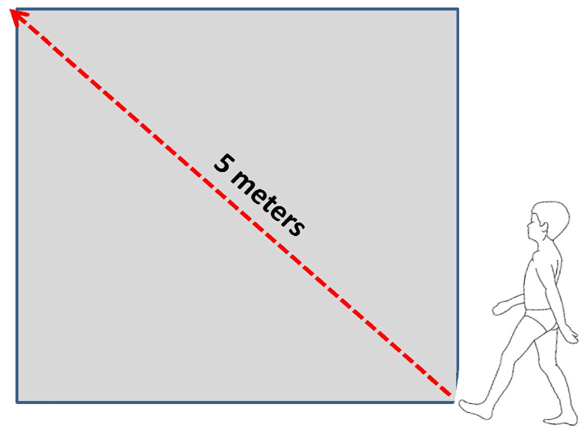


Figure 34. Walking area.

Under each garment condition, each subject repeated the walking test five times. For each walking test for each garment condition, the subject walked in a line from the designated start point to the end point. After completing a walking test, the subject returned to the start point to prepare the next walking test.

The walking test for each garment condition took about ten minutes including the time for system preparation and data saving. Each subject performed walking tests for each garment condition. The order of the garment conditions was set using a Latin Square Design (see Figure 33). Note each garment condition was evaluated once per subject and once per period.

To avoid fatigue caused by weight bearing, a resting time was given as follows. Two minutes of rest were allocated for treatments 1 and 2. Three minutes each were allocated after completing the walking test for treatments 3, 4 and 5, which each had a 20 lbs carrying load. Five minutes each was allocated after the walking tests for treatments 6 and 7, which each had a 40 lbs carrying load (see Figure 35). This part of the protocol was in total about 95 minutes.

T 1 + 2 min rest	T 2 + 2 min rest	T 3 + 3 min rest	T 4 + 3 min rest	T 5 + 3 min rest	T 6 + 5 min rest	T 7 + 5 min rest

Figure 35. Rest time for each garment condition.

During the rest time after performing the walking test for each garment condition, the subject relaxed with the carrying load removed. Then, the researcher asked the subject to indicate his perceived discomfort and fatigue level during the last completed walking test by orally responding to the researcher’s questions using a 5-point response scale on a prepared discomfort and fatigue ballot (Appendix F) and body area diagram (see Figure 36) identifying the body areas for which discomfort and/or fatigue was experienced. This ballot was modified from Nam’s (2009) perceived garment impediment ballot for a fit and performance evaluation of an arm armor system.

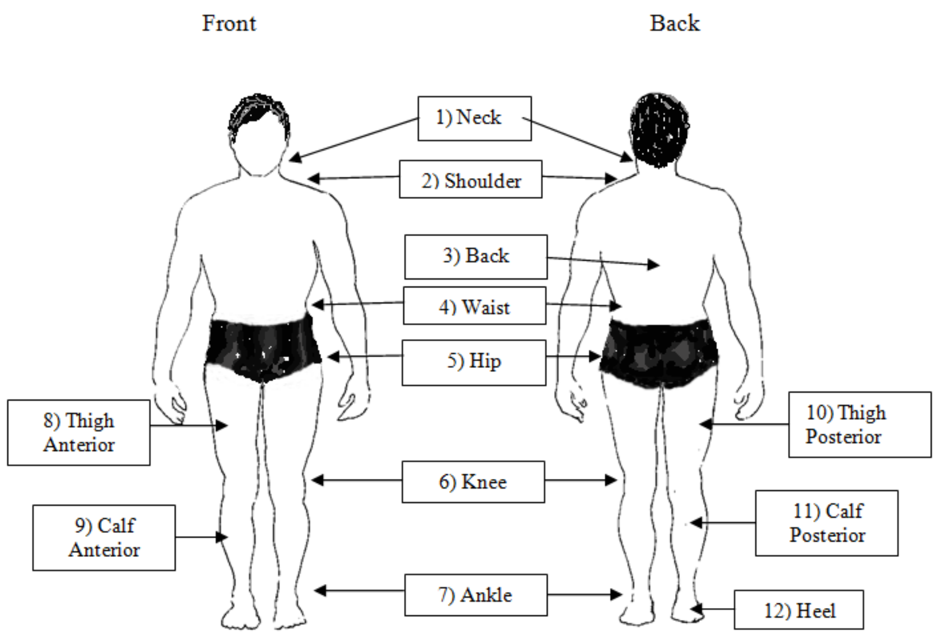


Figure 36. Diagram for discomfort and fatigue ballot.

Figure 37 shows the overall experiment protocol with approximate time per activity. Approximately three and a half hours were taken to complete all tests.

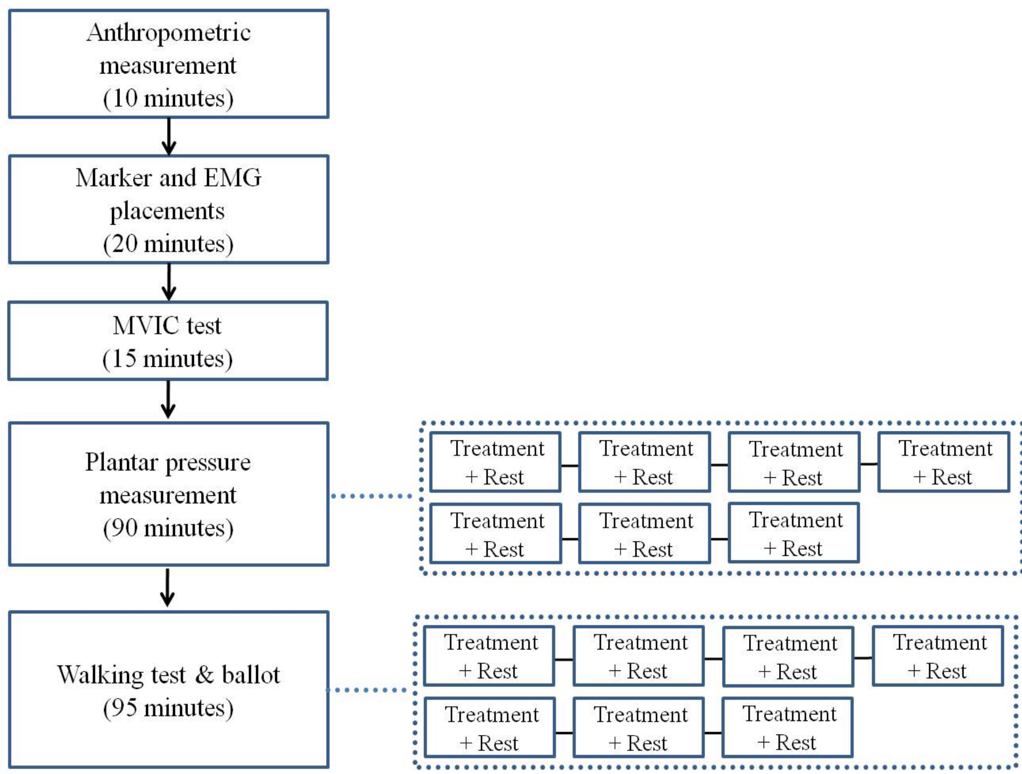


Figure 37. Experiment protocol.

Data Analysis

No subject reported an orthopedic disorder. However, one subject had a minor cartilage surgery on his right knee five years ago. All data collected from the subject were compared with normative band data and the data set collected from the other six subjects. Since the data were normal and there was no difference between the left and right sides, all data collected from the seven subjects were included in all data analyses.

A processing computer of the BTS system calculated dependent variables based on the Davis protocol. A total of 245 collected measurements (7 subjects x 7 garment conditions x 5 repetitions) from walking tests were analyzed to identify the effects of garment weight and weight distribution of carrying loads on ROM, temporal and distance parameters, EMG, plantar pressure and contact area. To analyze the change in plantar pressure, a total of 196 measurements (7 subjects x 7 garment conditions x 4 repetitions) were analyzed.

Measurements were first averaged on each side for each participant for each garment condition. A mixed models repeated measures analysis was performed using either the SAS/MIXED® procedure or SAS/GLIMMIX® procedure, Version 9.2 of the SAS System. Copyright © [2010] SAS Institute Inc. (Cary, NC, USA). The experiment design was a crossover design with repeated measures where participants and the order of testing garment conditions were blocks and side defined the repeated measures. Main effects of order, garment, and LR (left and right side) and the interactions were assessed. When garment effect was significant, post-hoc tests using Tukey pairwise comparisons and trend analyses were conducted. For the trend analysis, four levels of garment weight (1/8 lb: T1, 20 lbs: T2, 40 lbs: T3, T4 and T5, and 60 lbs: T6 and T7) were contrasted using orthogonal polynomial contrasts. Trend analysis was performed to identify the overall trend of responses to the garment weights. Due to the limited number of levels of garment weight, no regression equations were determined. All statistical tests were done at the .05 level of significance.

CHAPTER IV

RESULTS

Introduction

A mixed model repeated measures analysis using either the SAS/MIXED® procedure or SAS/GLIMMIX® procedure, was performed to determine a significant effect of the independent variables on the dependent variables of temporal and distance parameters, ROM, joint angle$_{max}$, EMG, plantar pressure and contact area, and interactions. When an effect was significant, Tukey pairwise post-hoc tests were performed to discern the mean difference(s) among garment treatments. A trend analysis was performed to identify any linear or curvilinear trends in the responses to weight. Four levels of garment weight (1/8 lbs: T1, 20 lbs: T2, 40 lbs: T3, T4 and T5, and 60 lbs: T6 and T7) were contrasted using orthogonal polynomial contrasts for the trend analysis. The perceptual data which assessed subjects' perceptions of wearing each garment were analyzed with qualitative analysis.

Temporal and Distance Parameters

Results of a repeated measures mixed model analysis for temporal and distance Parameters were summarized in Table 3, which included all least squares means and standard errors.

No significant interaction between garment and LR was observed for any of temporal or distance parameters. Least squares means and approximate trends are presented in Figure 38 – 41, where an approximate trend line of least squares means versus weight of carrying load is included.

Stance Phase

A significant garment effect was found (p = 0.0003) for stance phase (see Table 4).

Table 4

Temporal and Distance Parameter Least Squares Means, Standard Errors, and Significance Levels for Garment and LS Effects

Garment Effect	**Garment treatments**								
	T1	**T2**	**T3**	**T4**	**T5**	**T6**	**T7**		
Total weight (lb)	1/8	20	40	40	40	60	60		
Weight distribution									
	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	Std Error	p-value
***Stance Phase (%)**	64.54^{c}	65.16^{bc}	66.11^{a}	65.22^{abc}	65.29^{abc}	65.81^{ab}	65.47^{ab}	0.3696	**0.0003**
***Swing Phase (%)**	35.46^{a}	34.84^{ab}	33.89^{b}	34.78^{ab}	34.71^{ab}	34.197^{b}	34.53^{ab}	0.392	**0.0017**
***Double Support (%)**	14.49^{c}	15.07^{bc}	16.02^{a}	15.19^{bc}	15.41^{ab}	15.86^{ab}	15.49^{ab}	0.363	**<0.0001**
***Stride Length (m)**	1.32	1.31	1.26	1.25	1.27	1.27	1.27	0.0403	**0.0412**
Step Width (m)	0.17	0.2004	0.18	0.19	0.19	0.23	0.21	0.022	0.5321
Walking Speed (m/s)	1.06	1.06	0.999	1.01	1.02	1.03	0.98	0.024	0.0556

LR Effect	**Right**		**Left**		
	LSMean	Std Error	LSMean	Std Error	p-value
***Stance Phase (%)**	65.82	0.331	64.92	0.331	**<0.0001**
***Swing Phase (%)**	34.18	0.335	35.08	0.361	**<0.0001**

Note. Bold letter with * indicates a statistical significance for the effect.

abc means with a common superscripts in the same row are not significantly different (Tukey, α =0.05).

Stance phase was significantly longer ($p < 0.037$) while wearing T3 (66.11%) than while wearing T1 (64.54%) and T2 (65.16%). T6 (65.81%) and T7 (65.47%) stance phase means were significantly longer ($p \leq 0.0438$) than T1 (64.54%). A significant linear trend across four levels of garment weight ($p < 0.0001$) was found. Stance phase <u>increased</u> as the weight of garment and carrying load increased as shown in Figure 38. The right foot (65.82%) had a significantly longer stance phase than the left foot (64.92%) ($p < 0.0001$).

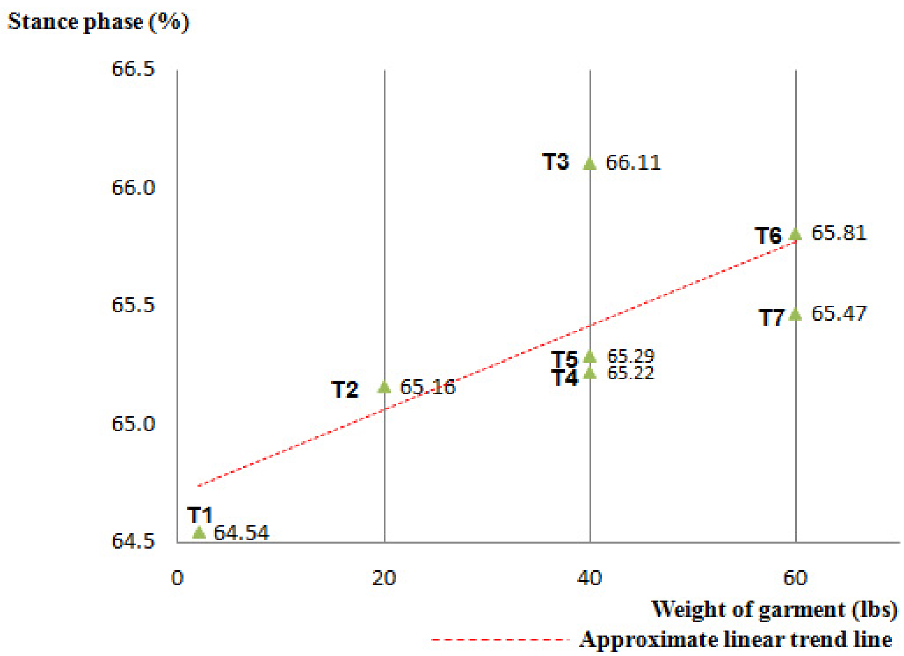


Figure 38. Stance phase least squares means and trend.

Swing Phase

A significant garment effect was found ($p = 0.0017$) for swing phase. Swing phase while wearing T3 (33.89%) and T6 (34.197%) was significantly shorter than while wearing T1 (35.46%) as shown in Table 4. A significant linear trend across four levels of weight of garment and carrying load was found ($p = 0.0003$). Swing phase significantly <u>decreased</u> with an increase in weight of garment and carrying load as shown in Figure 39. The right foot (34.18%) has a significantly shorter swing phase than the left foot (35.08%) ($p < 0.0001$).

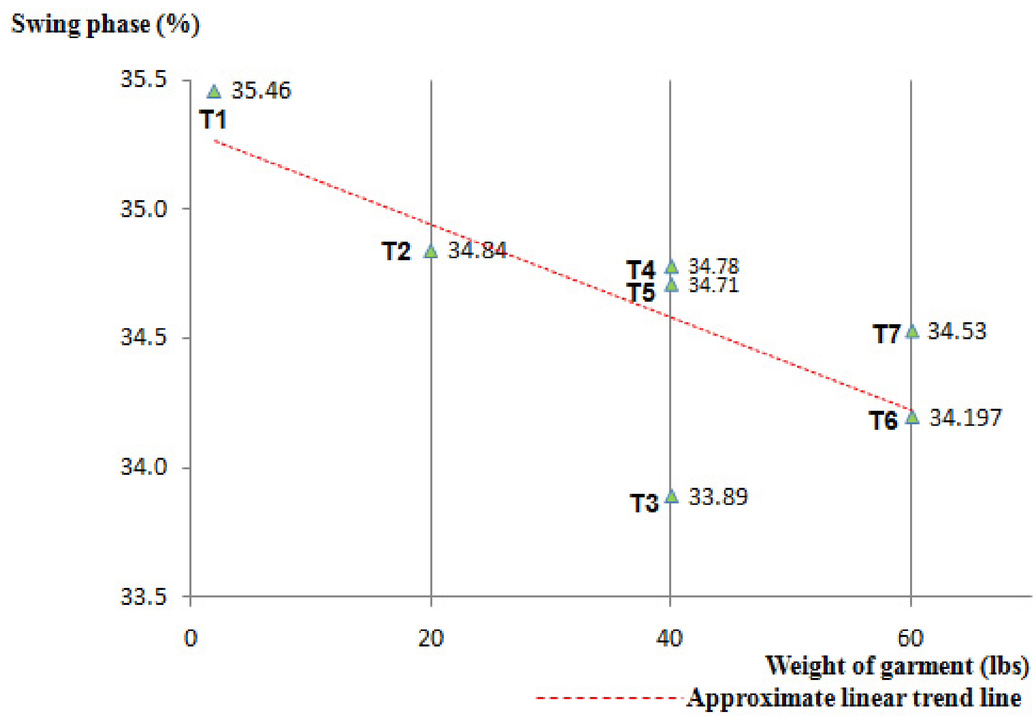


Figure 39. Swing phase least squares means and trend.

Double Support

A significant garment effect was found ($p < 0.0001$). Post-hoc tests supported that double support while wearing T3 (16.02%), T5 (15.41%), T6 (15.86%) and T7 (15.49%) was significantly longer than while wearing T1 (14.49%). In addition, double support while wearing T3 (16.02%) was significantly longer than double support while wearing T2 (15.07%) and T4 (15.19%). A significant linear trend across four levels of weight of garment and carrying load was found ($p< 0.0001$). Double support significantly <u>increased</u> as the weight of the garment and carrying load increased (see Figure 40).

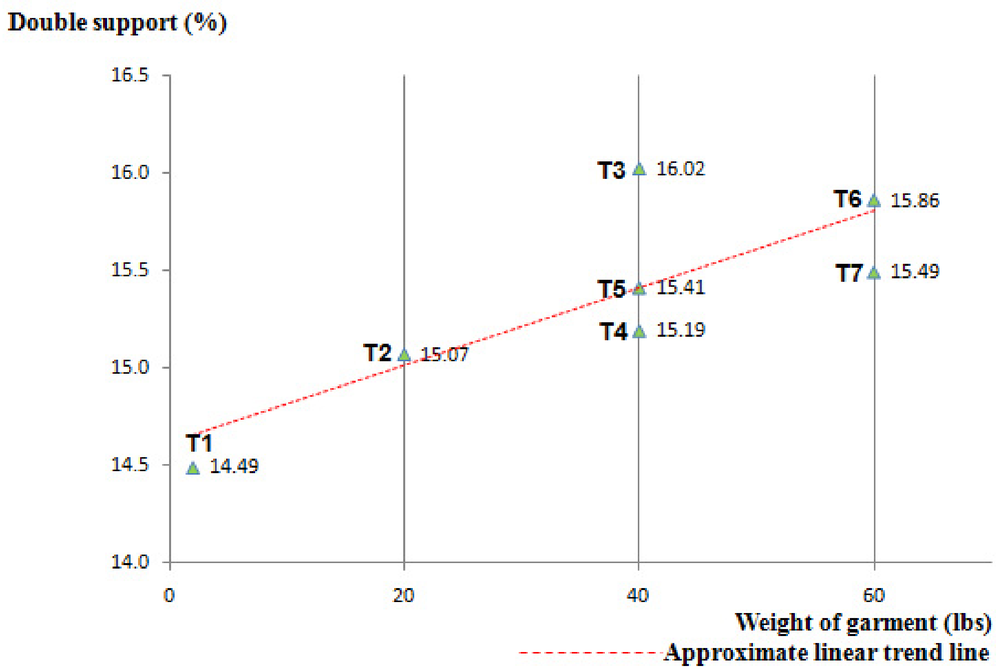


Figure 40. Double support least squares means and trend.

Stride Length

There was a significant garment effect (p = 0.0412). However, the conservative method of Tukey pairwise comparisons found no significant mean difference among garment treatments. A significant linear trend across four levels of weight of garment and carrying load was found (p = 0.0031) (see Figure 41). Stride length linearly <u>decreased</u> as the weight of the garment and carrying load increased.

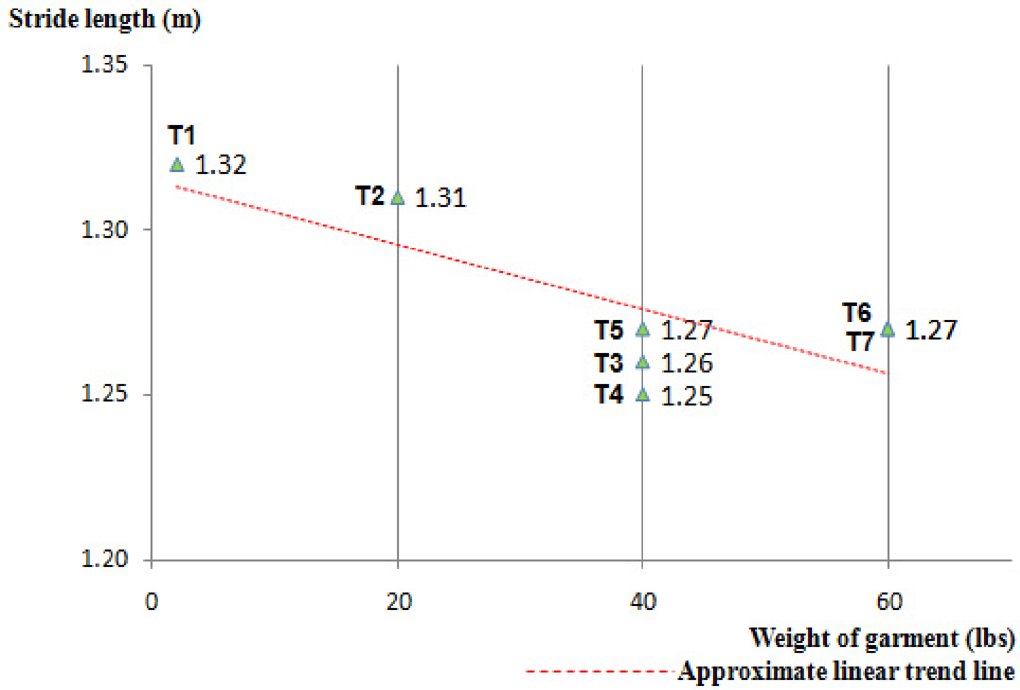


Figure 41. Stride length least squares means and trend.

No garment effect was found for step width nor walking speed.

Summary of Results for Temporal and Distance Parameters

A significant garment effect was found for stance phase, swing phase, double support and stride length. A significant garment effect and a linear trend support that stance phase and double support increased with an increase in the weight of garment and carrying load, while swing phase and stride length decreased. These results indicate that the foot contacts the floor for a longer period of time when the weight of the garment and carrying load increases.

Although a previous study (Park, Nolli, Branson, Peksoz, Petrova, and Goad, in press) found a significant difference between T1 and T2 for stance phase, swing phase and double support, the current study did not. The data pattern was consistent with the previous study in that stance phase and double support increased and swing phase decreased when subjects wore T2. However, the mean difference did not reach a statistical significance in this study. A possible reason for the difference may be the fitness of participants in both studies. Participants in this study were ROTC students who receive physical training on a regular basis, while participants in the previous study were students who may or may not have participated in physical training. Therefore, the impact of garment weight may be less on participants in this study than participants in the previous study.

The impact of weight distribution on walking patterns was particularly noticeable for the three 40 lb test garments (T3, T4 and T5). Significant results of stance phase and double support showed the mean for T3 (OTV + 20 lb carrying load on the left torso) was larger than T4 (OTV + 20 lb carrying load on the right torso) and T5 (OTV + each 10 lb on the left and right torso). This suggests that the placement of weight on the non-dominant side influenced these results. In particular, T3, which was manipulated to add the 20 lbs on the left torso, resulted in the longest foot contact time with the floor among the seven garment treatments as supported by the greatest stance phase and double support, and smallest swing phase.

No significant difference in walking patterns (stance phase, swing phase, double support and stride length) was found between T6 (OTV + 40 lb carrying load evenly distributed on the front and back torso) and T7 (OTV + 40 lb carrying load placed on the back).

A significantly longer stance phase and shorter swing phase found for the right foot suggests that the right-handed participants had a tendency to use the right leg more dominantly for weight bearing and maintaining body stability.

Joint Movement

To identify the effect of garment on joint movement, ROM and joint angle$_{max}$ were analyzed by using GLIMMIX® procedure. Then Tukey post-hoc pairwise comparison and trend analysis were further performed when a significant garment effect was found. No interaction between garment and LR was observed for any of ROM or joint angle$_{max}$. Least squares means and approximate trend were presented in Figure 42–49.

ROM

Significant garment effects were found for the pelvis and hip as shown in Table 5. Garment treatment with varying weight and weight distribution significantly influenced ROM for: pelvic obliquity, hip adduction-abduction, pelvic tilt, hip flexion-extension, pelvic rotation and hip rotation (see Table 5). A significant LR effect was found for knee rotation and hip rotation, which means participants' ROM for these two movements were significantly different between the left and right sides. Other than these two movements, there was no significant difference in ROM between the left and right sides (see Table 5). Detailed results are discussed in the following section.

Frontal Plane Movements

Pelvic obliquity: There was a significant garment effect ($p = 0.0022$). Post-hoc tests showed a significant difference in ROM for pelvic obliquity between (T1 and T3), (T1 and T5), and (T2 and T3) as shown in Table 5 ($p \leq 0.0405$). Overall, the ROM for pelvic obliquity tended to <u>decrease</u> with an increase in the weight of garment and carrying loads. A significant linear trend ($p = 0.0004$) across four levels of weight of garment and carrying load was found as shown in Figure 42.

Table 5

ROM Least Squares Means, Standard Errors, and Significance Levels for Garment and LR Effects

Garment Effect	**Garment treatments**								
	T1	**T2**	**T3**	**T4**	**T5**	**T6**	**T7**		
Total weight (lb)	1/8	20	40	40	40	60	60		
Weight distribution									
	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	Std Error	p-value
Frontal Plane Movement									
***Pelvic obliquity**	6.16^{a}	6.08^{ab}	4.48^{c}	4.87^{abc}	4.65^{bc}	4.74^{abc}	5.16^{abc}	0.515	**0.0022**
***Hip adduction-abduction**	13.37^{a}	13.197^{ab}	11.78^{b}	12.19^{ab}	12.3^{ab}	12.89^{ab}	13.11^{ab}	0.646	**0.0239**
Sagittal Plane Movement									
***Pelvic tilt**	3.36^{b}	3.34^{b}	3.404^{b}	3.35^{b}	3.19^{b}	3.05^{b}	4.61^{a}	0.198	**< .0001**
***Hip flexion-extension**	40.99^{b}	42.58^{ab}	41.44^{ab}	40.89^{b}	41.704^{ab}	42.99^{ab}	44.86^{a}	1.504	**0.0272**
Knee flexion	58.24	58.46	57.90	57.77	58.27	58.46	57.62	1.69	0.97
Ankle flexion	30.44	31.46	30.42	29.22	30.39	30.19	30.54	1.74	0.13
Transverse Plane Movement									
***Pelvic rotation**	9.56^{a}	6.77^{b}	7.5007^{ab}	6.98^{b}	6.396^{b}	6.48^{b}	5.94^{b}	0.904	**0.0027**
***Hip rotation**	14.04^{ab}	13.37^{ab}	12.18^{b}	12.52^{b}	12.74^{b}	13.52^{ab}	14.82^{a}	0.97	**0.0062**
Knee rotation	23.51	24.97	24.71	22.86	23.44	24.04	23.24	1.78	0.1577
Ankle rotation	15.73	16.101	15.27	14.79	15.63	16.68	16.44	1.74	0.13

LR Effect	**Right**		**Left**		
	LSMean	Std Error	LSMean	Std Error	p-value
***Pelvic tilt**	3.36	0.12	3.58	0.15	**0.0301**
***Pelvic rotation**	7.389	0.76	6.789	0.76	**< .0001**
***Hip rotation**	11.97	0.899	14.66	0.93	**< .0001**
***Knee rotation**	25.13	1.725	22.52	1.76	**0.0044**

Note. Bold letter with * indicates a statistical significance for the effect.

^{abc} means with a common superscripts in the same row are not significantly different (Tukey, α =0.05).

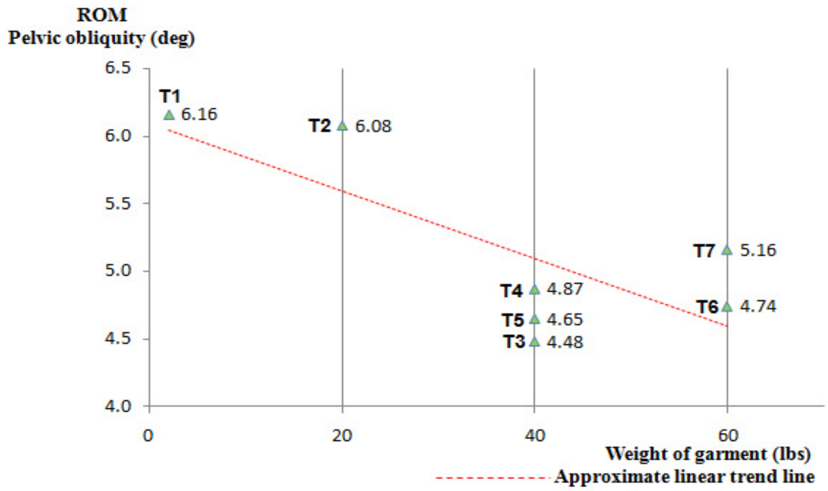


Figure 42. ROM of pelvic obliquity least squares means and trend.

Hip adduction-abduction: There was a significant garment effect (p = 0.0239). Post-hoc tests showed that the ROM for hip adduction-abduction (p = 0.0457) was significantly smaller while wearing T3 (11.78 deg) than while wearing T1 (13.37 deg). There was no significant linear trend of the decreased ROM for hip adduction-abduction (see Figure 43).

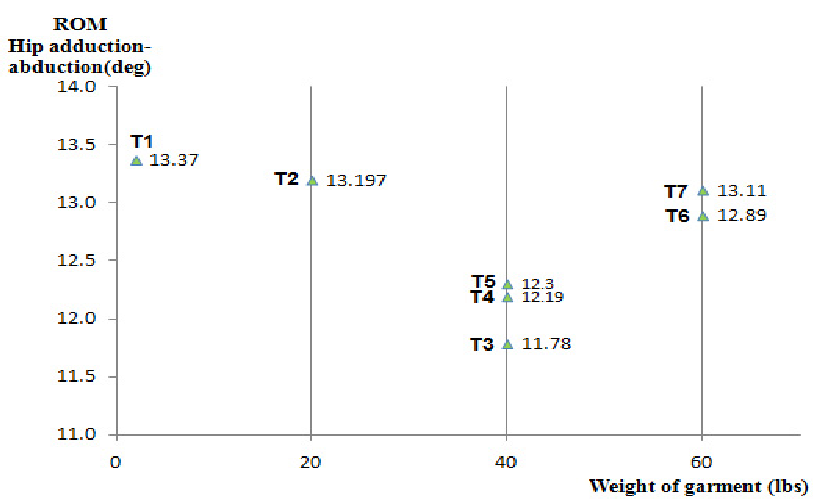


Figure 43. ROM of hip adduction-abduction least squares means.

Sagittal Plane Movements

Pelvic tilt: There was a significant garment effect ($p < 0.0001$). Post-hoc tests showed that ROM for the pelvic tilt was significantly larger while wearing T7 (4.61 deg) than while wearing all other six treatments ($p \leq 0.0006$). Overall, ROM for pelvic tilt tended to <u>increase</u> with an increase in weight of the garment and carrying loads. A significant linear trend was found across four levels of weight of garment and carrying load ($p = 0.047$) as shown in Figure 44. ROM was significantly greater on the left side (3.58 deg) than on the right side (3.36 deg) ($p = 0.0301$).

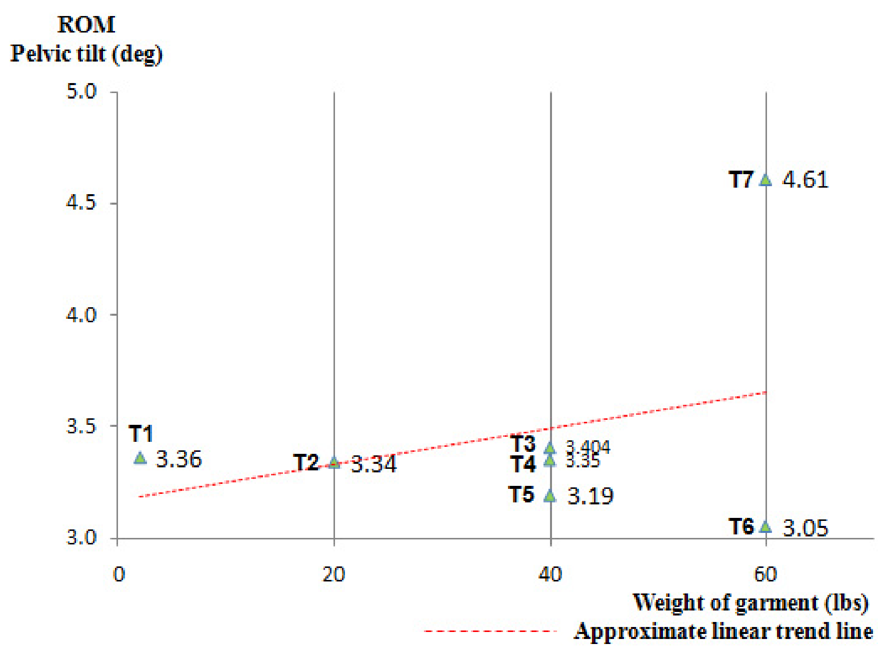


Figure 44. ROM of pelvic tilt least squares means and trend.

Hip flexion-extension: There was a significant garment effect ($p = 0.0272$). Post-hoc tests showed that ROM for hip flexion-extension was significantly larger while wearing T7 (44.86 deg) than while wearing T1 (40.99 deg) and T4 (40.89 deg). There was a linear trend of increased

ROM for the hip flexion-extension across four levels of weight of garment and carrying load (p = 0.0252) as shown in Figure 45.

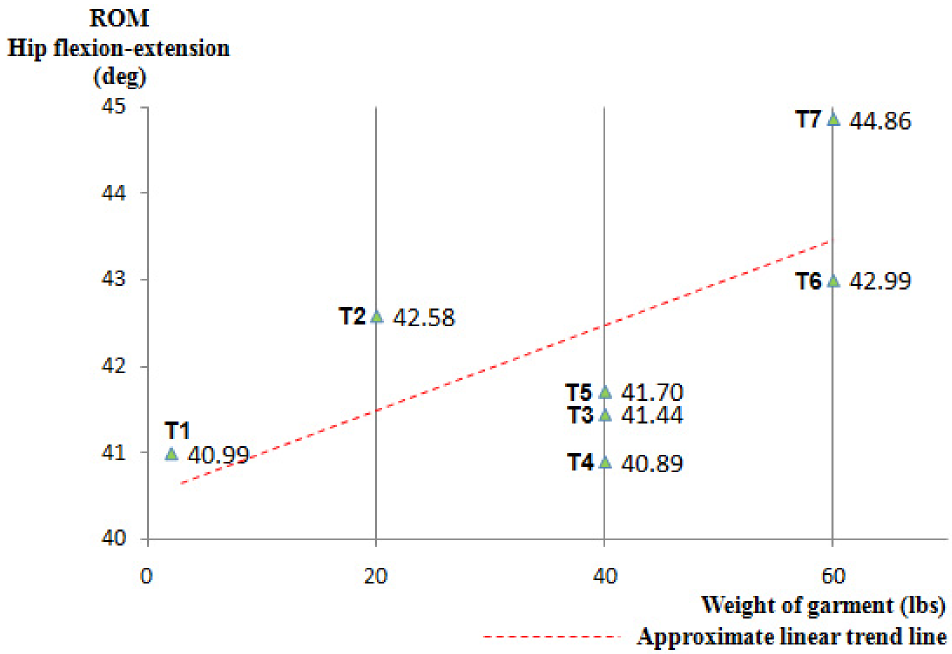


Figure 45. ROM of hip flexion-extension least squares means.

Knee flexion & ankle flexion: There was no garment effect nor LR effect.

Transverse Plane Movements

Pelvic rotation: There was a significant garment effect (p = 0.0027). Post-hoc tests showed that ROM for pelvic rotation was significantly larger while wearing T1 (9.56 deg) than while wearing T2 (6.77 deg), T4 (6.98 deg), T5 (6.396 deg), T6 (6.48 deg) and T7 (5.94 deg) as shown in Table 5. ROM for pelvic rotation was greater on the right side (7.389 deg) than on the left side (6.789 deg) ($p < 0.001$).

Overall, ROM for pelvic rotation decreased as the weight of garment and carrying loads increased. A significant linear trend (p = 0.0001) across four levels of weight of garment and carrying load was found by a trend analysis (see Figure 46).

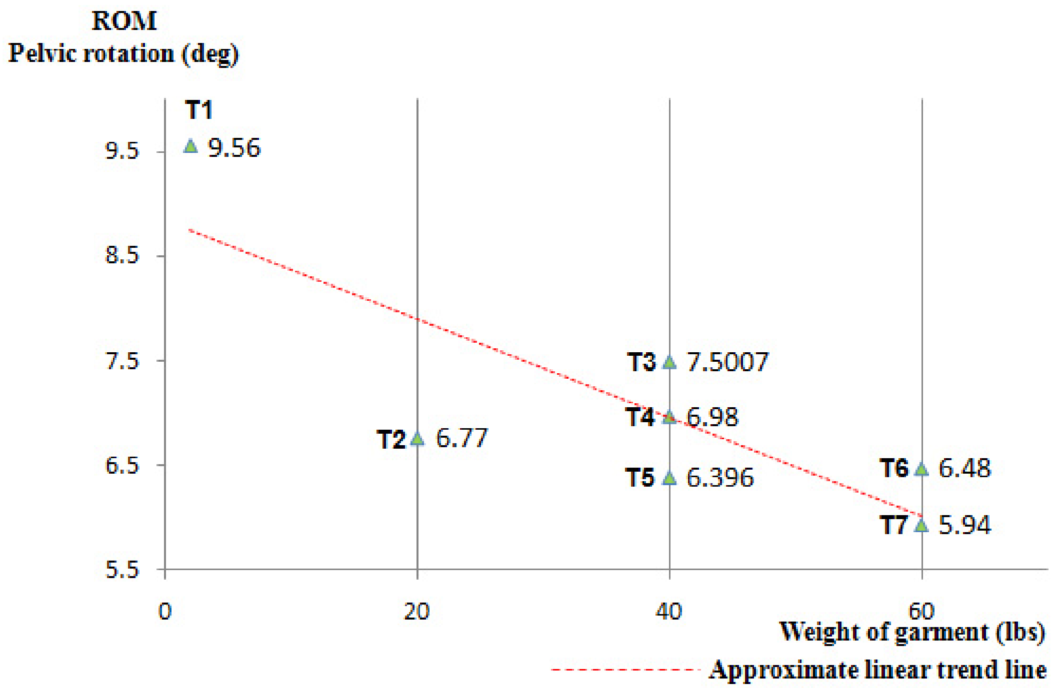


Figure 46. ROM of pelvic rotation least squares means and trend.

Hip rotation: A significant garment effect was found for the hip rotation ROM (p = 0.0062). Post-hoc tests showed a significant mean difference in ROM for hip rotation; ROM was significantly larger while wearing T7 (14.82 deg) than while wearing T3 (12.18 deg), T4 (12.52 deg) and T5 (12.74 deg) as shown in Table 5. The trend of ROM data for hip rotation was not linear with an increase in weight magnitude (see Figure 47). Subjects showed significantly larger ROM on the left side (14.66 deg) than on the right side (11.97 deg) ($p < 0.0001$).

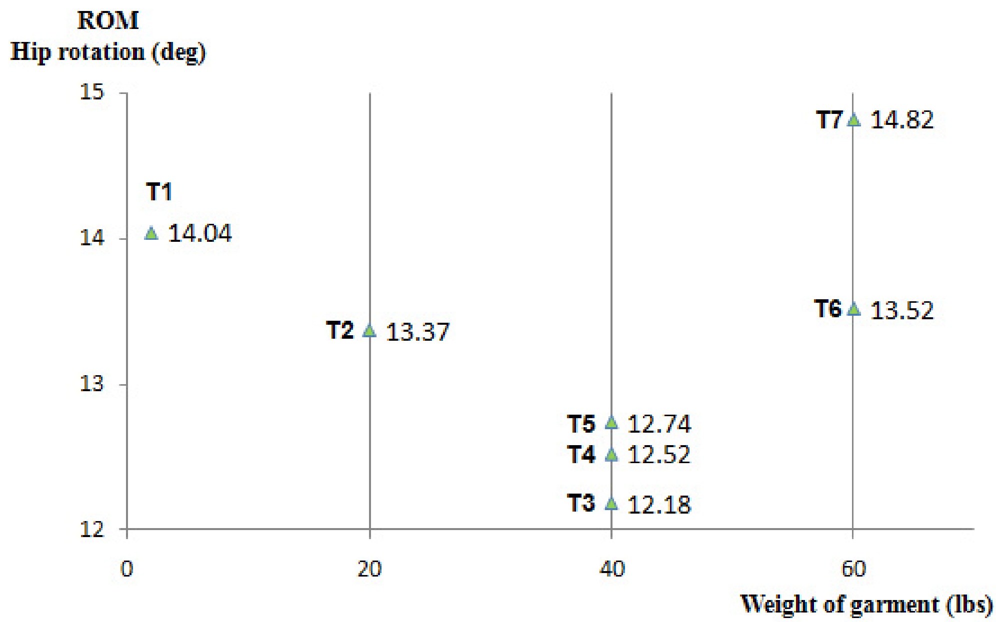


Figure 47. ROM of hip rotation least squares means.

Knee rotation: There was no garment effect. ROM was significantly greater on the right side (25.13 deg) than on the left side (22.52 deg) ($p = 0.004$).

Ankle rotation: There was no garment effect nor LR effect. A significant order effect was found for the ankle rotation ($p = 0.0161$).

Joint Angles$_{max}$

Significant garment effects were found only for intra- and extra pelvic rotation. A significant LR effect was found for upward pelvic obliquity and downward pelvic obliquity, hip adduction, anterior pelvic tilt, hip flexion and ankle extrarotation. Therefore, subjects' joint angles$_{max}$ were significantly different between the left and right sides in these directional movements. Detailed results are presented in the following section.

Frontal Plane Movements

Upward / downward pelvic obliquity: There was no significant garment effect in the upward nor downward directions. A significant LR effect was found for upward pelvic obliquity ($p = 0.0002$) and downward pelvic obliquity ($p = 0.01$); Joint angle$_{max}$ for upward pelvic obliquity was greater on the right side (3.25 deg) than on the left side (2.29 deg). The absolute value of the joint angle$_{max}$ for the downward pelvic obliquity was greater on the left side (-3.198 deg) than on the right side (-2.48 deg) as shown in Table 6.

Hip adduction / abduction: There was no significant garment effect. The joint angle$_{max}$ for hip adduction was significantly greater on the right side (1.315 deg) than on the left side (0.716 deg) ($p = 0.0278$).

Sagittal Plane Movements

Pelvic tilt: Posterior pelvic tilt was not included in the statistical analysis because five subjects did not show posterior pelvic tilt, which is considered normal according to the literature (Gage, 2004: Leardini et al., 2007). Therefore, only anterior pelvic tilt data were analyzed. There was no significant garment effect for anterior pelvic tilt. The joint angle$_{max}$ for anterior pelvic tilt was significantly greater on the left side (6.88 deg) than on the right side (6.68 deg) ($p = 0.004$).

Hip flexion-extension: There was no significant garment effect for joint angle$_{max}$ for hip flexion and extension. The joint angle$_{max}$ for hip flexion was significantly greater on the left side (36.268 deg) than on the right side (35.302 deg) ($p = 0.0012$). A significant order effect was found for hip flexion ($p = 0.0305$).

Ankle dorsiflexion-plantarflexion: There was no significant effect of garment nor LR on joint angle$_{max}$ for ankle dorsiflexion and plantarflexion.

Table 6

Joint Angle$_{max}$ Least Squares Means, Standard Errors, and Significance Levels for Garment and LR Effects

Garment Effect	**Garment treatments**								
	T1	**T2**	**T3**	**T4**	**T5**	**T6**	**T7**		
Total weight (lb)	1/8	20	40	40	40	60	60		
Weight distribution									
	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	Std Error	p-value
Frontal Plane Movement									
Pelvic obliquity - Upward	3.02	3.09	2.51	2.81	2.66	2.57	2.74	0.30	0.811
Pelvic obliquity - Downward	-3.14	-3.01	-2.44	-3.04	-2.76	-2.80	-2.69	0.34	0.798
Hip adduction	1.15	0.62	1.08	1.18	1.17	0.66	1.25	0.203	0.303
Hip abduction	-13.87	-13.68	-13.23	-13.42	-13.47	-13.76	-13.88	0.54	0.653
Sagittal Plane Movement									
Pelvic tilt - Anterior	7.64	6.63	7.38	8.27	5.87	6.08	5.61	1.21	0.372
Hip flexion	34.81	34.696	36.11	37.44	36.85	36.01	34.59	2.12	0.400
Hip extension	-7.11	-9.14	-7.12	-4.77	-5.93	-8.32	-10.48	2.79	0.195
Knee flexion	68.31	68.18	69.05	69.97	70.17	69.47	67.74	1.05	0.512
Ankle dorsiflexion	13.17	13.34	13.96	13.45	14.296	12.98	14.84	0.15	0.174
Ankle plantarflexion	-17.28	-18.12	-16.43	-15.77	-16.09	-17.24	-15.703	0.21	0.180
Transverse Plane Movement									
***Pelvic intrarotation**	4.87^{a}	3.77^{ab}	3.79^{ab}	3.397^{b}	3.43^{ab}	3.62^{ab}	3.44^{ab}	0.43	**0.048**
***Pelvic extrarotation**	-4.68^{b}	-3.55^{ab}	-4.13^{ab}	-4.004^{ab}	-3.52^{ab}	-3.76^{ab}	-3.21^{a}	0.45	**0.017**
Hip intrarotation	10.16	11.94	11.15	9.54	10.38	11.13	11.67	2.23	0.607
Hip extrarotation	-4.23	-5.47	-4.71	-5.49	-5.05	-5.66	-5.95	1.64	0.684
Knee intrarotation	15.09	15.07	15.55	15.77	16.05	16.03	16.92	1.54	0.376
Knee extrarotation	-10.099	-11.47	-9.77	-7.35	-8.63	-9.57	-7.605	2.31	0.078
Ankle extrarotation	-19.71	-19.82	-19.63	-20.098	-20.19	-19.29	-19.17	2.39	0.839

LR Effect	**Right**		**Left**		
	LSMean	Std Error	LSMean	Std Error	p-value
***Pelvic obliquity - Upward**	3.25	0.17	2.29	0.18	**0.0002**
***Pelvic obliquity - Downward**	-2.48	0.22	-3.198	0.21	**0.010**
***Hip adduction**	1.315	0.19	0.716	0.14	**0.0278**
***Pelvic tilt - Anterior**	6.68	0.86	6.88	0.86	**0.004**
***Hip flexion**	35.302	1.897	36.268	1.867	**0.0012**
***Ankle extrarotation**	-20.803	0.001	-18.6007	0.003	**0.0003**

Note. Bold letter with * indicates a statistical significance for the effect.

^{abc} means with a common superscripts in the same row are not significantly different (Tukey, α =0.05).

Transverse Plane Movements

Pelvic intrarotation – extrarotation: A significant garment effect was found for pelvic intrarotation (p = 0.048) and extrarotation (p = 0.017). Post-hoc tests for pelvic intrarotation showed a significant mean difference between T1 and T4 as shown in Table 6. Pelvic intrarotation was significantly smaller in T4 (3.397 deg) than in T1 (4.87 deg) as shown in Table 6. Overall, pelvic intrarotation decreased with an increase in weight of the garment and carrying loads. In pelvic intrarotation, a significant linear trend was found across four levels of weight of garment and carrying load (p = 0.0041) as shown in Figure 48.

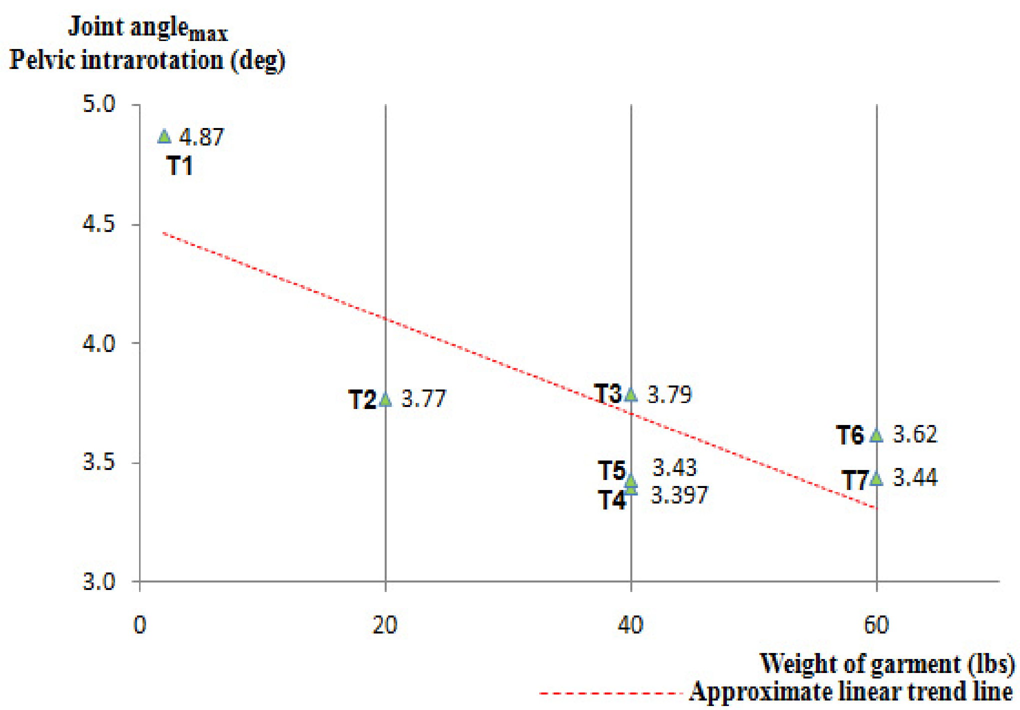


Figure 48. Joint angle$_{max}$ for pelvic intrarotation least squares means and trend.

Joint angle for pelvic extrarotation had negative values because the movement direction is opposite to the direction of intrarotation as explained in Table 2. Extrarotation at each joint has negative values while intrarotation has positive values, which indicates the direction of movement.

Post-hoc tests showed that the absolute value of the pelvic extrarotation was significantly smaller while wearing T2 (-3.55 deg) and T7 (-3.21 deg) than while wearing T1 (-4.68 deg). There was no significant LR effect. Pelvic extrarotation also showed a significant linear trend ($p = 0.0045$) across four levels of weight of garment and carrying load as shown in Figure 49. Overall, extrarotation decreased with an increase in weight of the garment and carrying loads.

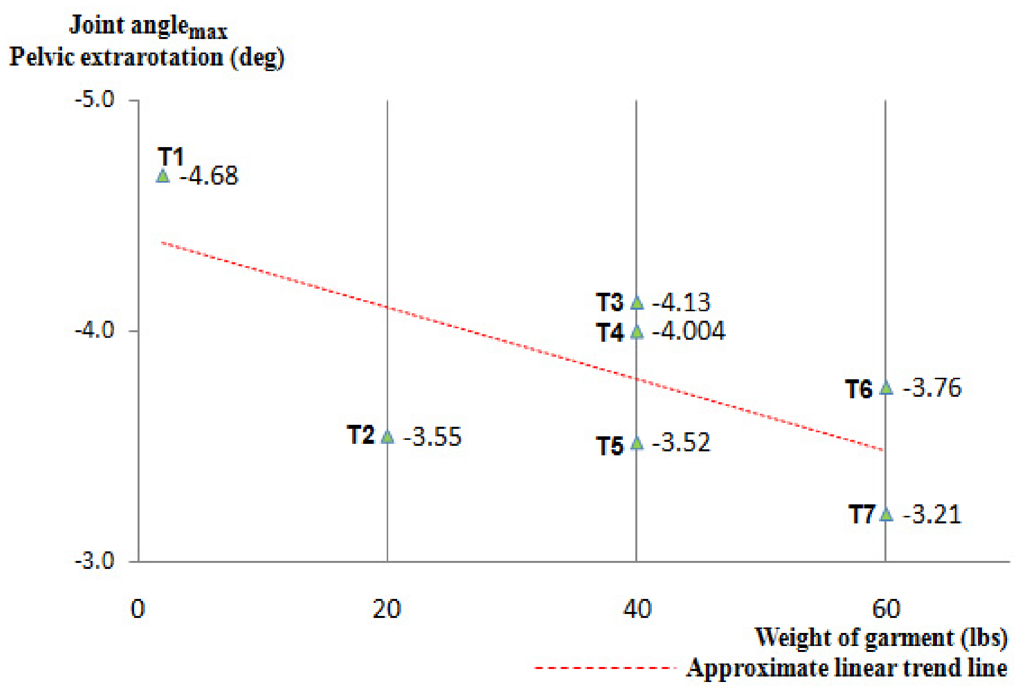


Figure 49. Joint angle$_{max}$ for pelvic extrarotation least squares means and trend.

Hip intrarotation – extrarotation: There was no significant garment effect nor LR effect for hip intrarotation and extrarotation. A significant order effect was found for hip rotation ($p = 0.0497$).

Knee intrarotation – extrarotation: There was no significant garment effect nor LR effect for knee intrarotation and extrarotation.

Ankle intrarotation-extrarotation: Ankle intrarotation was not included in the statistical analysis because all seven subjects did not show ankle intrarotation, which is considered normal according to the literature (Gage, 2004: Leardini et al., 2007). Therefore, only ankle extrarotation was analyzed. There was no significant garment effect for extrarotation. A significant LR effect was found for the joint $angle_{max}$ for extrarotation ($p = 0.0003$). The right foot (-20.803 deg) showed a significantly larger absolute value of the joint $angle_{max}$ than the left foot (-18.6007 deg).

Summary of Results for ROM and Joint $Angle_{max}$

Garment treatment significantly influenced ROM in the pelvis and hip. Pelvic movement appears to be affected by the level of garment weight as evidenced by a significant linear trend found for pelvic obliquity, rotation and tilt; ROM for pelvic obliquity and pelvic rotation significantly decreased with an increase in the weight of garment and carrying loads. ROM for pelvic tilt significantly increased with an increase in the weight of the garment and carrying loads. In particular, while wearing T7 (OTV + 40 lb carrying load placed on the back), subjects showed a significantly larger ROM for pelvic tilt than while wearing T6 (OTV + 40 lb carrying load evenly distributed on the front and back torso). This result supports the impact of weight distribution on the pelvic tilt while walking. An increase in weight is known to increase the forward lean of the trunk, which has been shown to increase ROM for anterior pelvic tilt (Smith et al., 2006). In this study, the greatest pelvic tilt found for T7, appears to result from placement of the greatest carrying load on the back, which may create the greatest forward lean of the trunk. On the other hand, T6 with uniform weight distribution around the torso showed the lower ROM for pelvic tilt than T7 despite the same 60 lb carrying load. Pelvic tilt while wearing T6 was statistically the same as pelvic tilt while wearing T1, T2, T3, T4 and T5.

With regard to hip movement, significant garment effects were found for ROM for hip adduction-abduction, hip flexion-extension and hip rotation. However, a linear trend was not significant in ROM for these hip movements. As weight of the garment and carrying load increased, ROM for hip flexion-extension increased with the exception of the three 40 lb garment treatments (T3, T4 and T5). Post-hoc tests for hip flexion-extension showed a significant difference in ROM between (T1 and T7) and (T4 and T7). ROM for hip adduction- abduction and hip rotation tended to decrease with increasing weight of garment and carrying loads. Significant LR effects were found for ROM for knee rotation, pelvic tilt and hip rotation.

Significant garment effects were found for joint angle$_{max}$ for pelvic intrarotation and extrarotation. Significant LR effects were found for joint angle$_{max}$ for upward/downward pelvic obliquity, hip adduction, anterior pelvic tilt, hip flexion and ankle extrarotation.

EMG

The garment effect on EMG was analyzed using GLIMMIX procedure by comparing maximum value of the normalized EMG amplitude while wearing each garment treatment. A significant garment effect was found for only rectus femoris. A significant LR effect was found for the bicep femoris long head and medial gastrocnemius, indicating that the normalized EMG was significantly different for the left and the right sides. No interaction between garment and LR was observed for any of EMG. Higher amplitude of muscle EMG indicates greater muscle activity to generate greater physical force. Detailed results are presented in the following section.

Rectus femoris: A significant garment effect was found (p = 0.0075) as shown in Table 7.

Table 7

Normalized EMG Least Squares Means, Standard Errors, and Significance Levels for Garment and LR Effects

Garment Effect	**Garment treatments**								
	T1	**T2**	**T3**	**T4**	**T5**	**T6**	**T7**		
Total weight (lb)	1/8	20	40	40	40	60	60		
Weight distribution									
	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	Std Error	p-value
***Rectus Femoris (%)**	1.93^{b}	2.04^{ab}	2.001^{b}	2.18^{ab}	2.11^{ab}	2.39^{ab}	2.53^{a}	0.22	**0.0075**
Bicep Femoris Long Head (%)	3.39	3.42	3.58	3.62	3.72	3.55	3.14	0.49	0.4396
Tibialis Anterior (%)	4.78	5.36	5.02	5.13	5.65	5.84	4.790	0.68	0.6819
Medial Gastrocnemius (%)	6.69	7.54	7.21	7.07	7.73	7.87	8.17	1.03	0.5814

LR Effect	**Right**		**Left**		
	LSMean	Std Error	LSMean	Std Error	p-value
***Bicep Femoris Long Head (%)**	2.97	0.47	4.01	0.49	**0.0002**
***Medial Gastrocnemius (%)**	6.84	0.86	8.10	0.95	**0.0057**

Note. Bold letter with * indicates a statistical significance for the effect.

^{abc} means with a common superscripts in the same row are not significantly different (Tukey, α =0.05).

Post-hoc tests showed a significant difference in the normalized EMG amplitude between (T1 and T7) and (T3 and T7) as shown in Figure 50; the normalized EMG amplitude was greater while wearing T7 (2.53%) than while wearing T1 (1.93%) and T3 (2.001%). During the stance phase, the normalized amplitude increased with an increase in the weight of garment and carrying load, while the normalized amplitudes during the swing phase were similar as shown in Figure 50.

A significant linear trend (p = 0.0007) was found across four levels of weight of garment and carrying load; Overall, the normalized EMG amplitude for rectus femoris increased with an increase in weight of the garment and carrying loads (see Figure 50). No significant difference in normalized amplitude between the left and right foot was found (see Table 7).

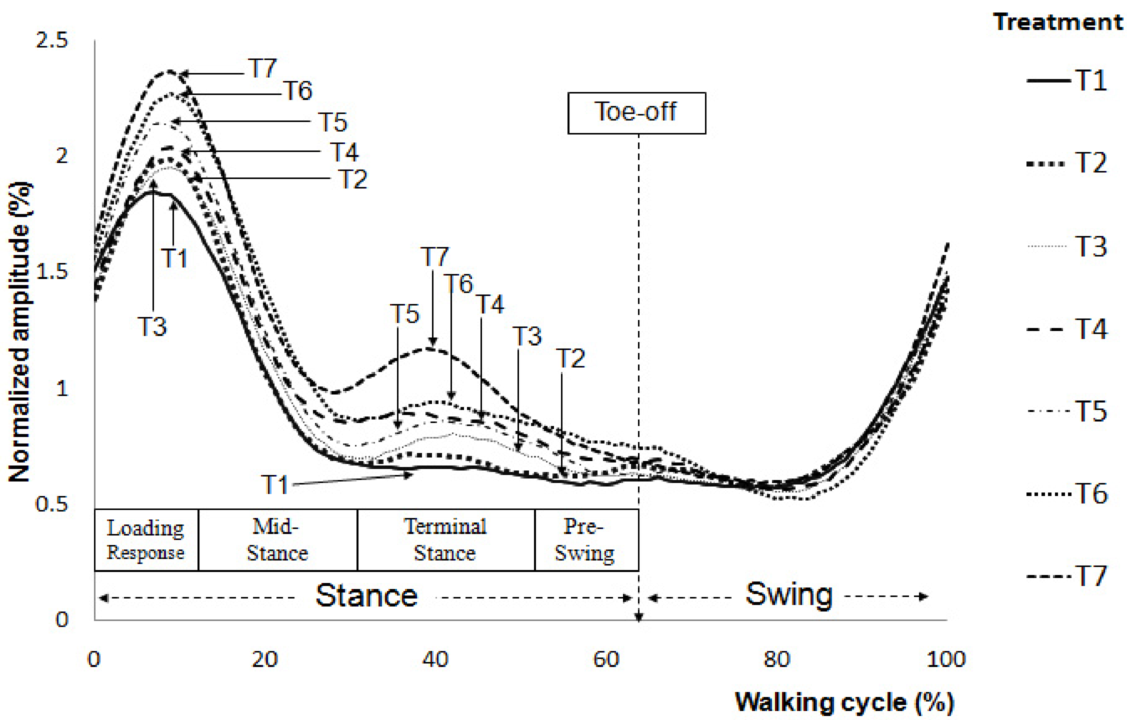


Figure 50. The normalized EMG on the right rectus femoris.

Bicep femoris long head: There was no significant garment effect. A significant LR effect was found (p = 0.0002); the normalized amplitude of EMG in the bicep femoris long head was greater on the left side (4.01%) than on the right side (2.97%) as shown in Table 7.

Tibialis anterior: There was no significant garment effect nor LR effect.

Medial gastrocnemius: There was no garment effect. The normalized amplitude of EMG was significantly greater on the left (8.10%) than on the right (6.84%) (see Table 7).

A significant garment effect was found for the EMG for rectus femoris. The maximum amplitude (the first peak in Figure 50) was found around 10% of the gait cycle. This is considered a normal EMG pattern for the rectus femoris according to the literature (Vaughan et al., 1996). In normal walking, the stance phase includes five subsequent events: initial contact (heel strike: 0%

of gait cycle), loading response (0-10% of gait cycle), mid-stance (10-30% of gait cycle), terminal swing (30-50% of gait cycle) and pre-swing (50-60% of gait cycle) as shown in Figure 51.

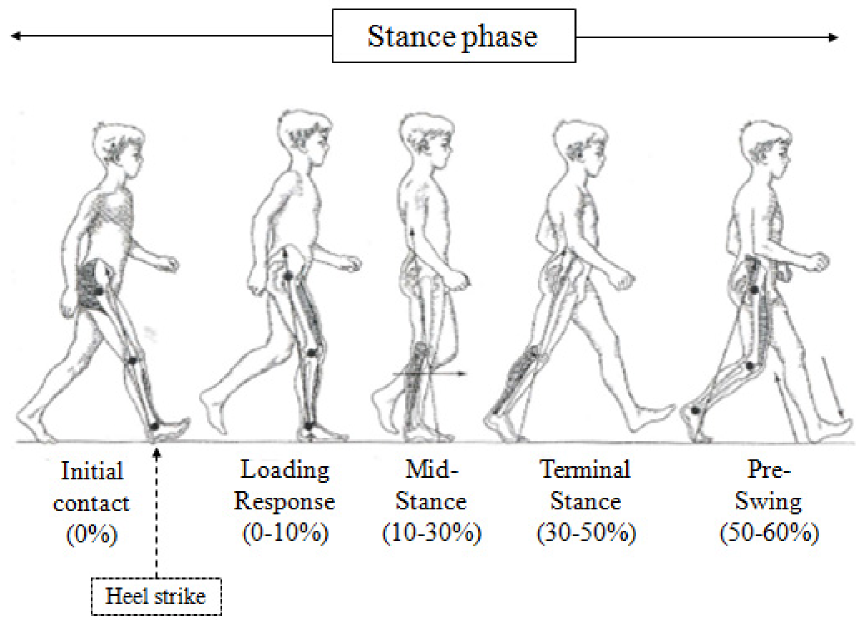


Figure 51. Sequential events of stance phase. Adapted from *The Treatment of Gait Problems in Cerebral Palsy*, (p. 52), by J. R. Gage, 2004, London, UK: Mac Keith Press.

During the loading phase, rectus femoris, as a hip and thigh stabilizer, works through eccentric muscle contraction, which is a type of muscle contraction by lengthening muscle fibers used as a means of decelerating a body part for stable movement. The eccentric muscle contraction of rectus femoris smoothly decelerates the inertia of the body moving forward and absorbs the impact forces by assisting with hip flexion and knee extension and decelerating knee flexion at the same time (Gage, 2004; Vaughan et al., 1996). Therefore, rectus femoris shows the peak amplitude to generate the force for hip flexion and knee extension (Vaughan et al., 1996). In this study, an increase in garment weight results in higher peak EMG in rectus femoris as supported by a significant linear trend across four levels of weight of garment and carrying load. A significant increase in amplitude during the loading phase implies that an increase in garment weight increases inertia of the lower limb for forward movement and the impact force and at the moment of heel strike (Gage, 2004), which may require rectus femoris to generate greater force to decelerate the lower limb, which may be necessary for maintaining body balance. In addition, a noticeable difference in the EMG signal among the garment treatments was also observed between 30% and 60% of the cycle, which is the terminal stance (30% - 50% of the gait cycle) and pre-swing (50-60% of the gait cycle). According to the literature, in the normal walking pattern as for T1 condition, the amplitude of the rectus femoris gradually decreases as the knee begins to flex during the mid-stance phase (10% - 30% of the gait cycle), Then the rectus femoris remains inactive until the end of the swing when the foot goes to the next heel strike, where hip flexion and knee extension are about to occur. However, in this study, in the case of T2 through T7 as shown in Figure 50, the second curve of the EMG signal was observed during the terminal stance (30-50%) and pre-swing (50-60%). Terminal stance is the period of time when the stance foot (lead foot)'s heel rises and the contralateral foot (trailing foot) contacts the ground. Pre-swing indicates the period of time when the stance foot is about to go to swing phase with toe off and the contralateral foot goes to the loading phase. During pre-swing, the body weight and inertia move quickly forward (Gage, 2004). Therefore, during the terminal stance and pre-swing (about 30- 60% of the gait cycle) at the last stage of stance phase, the body's center of gravity moves from the stance foot to the contralateral foot, which is about to enter stance phase. Therefore, in this study, the abnormal second EMG curve observed when subjects wore T2 through T7, during the terminal and pre-swing phase when the body's center of gravity shifts from one leg to the other, implies that the magnitude of garment weight increases the activity of the rectus femoris to maintain body balance and stable walking under weight-bearing condition.

Post-hoc tests supported that T7 with the 40 lb carrying load placed on the back created the higher peak muscle amplitude for rectus femoris, which showed greater muscle activity to generate greater force for the muscle.

Plantar Pressure and Contact Area

The effect of garment on 1) peak plantar pressure (PPP), 2) average plantar pressure on the sole of the foot, and 3) the change in contact area were analyzed. The change in contact area for T2 through T7 was calculated based on the formula shown below: contact area for each treatment (T #) during each test was subtracted from the mean contact area for T1, and then the calculated value was divided by the mean contact area for T1. The calculated change in contact area was expressed as a percent. Therefore, the contact area at T1 was used as a baseline. This manipulation was done to investigate change in contact area for garment treatment with no influence for individual differences in foot size.

$$\Delta \text{ Contact area (\%)} = \left[\frac{\text{Contact area (cm}^2\text{) at T \#}}{\text{Contact area (cm}^2\text{) at T1}} - 1 \right] \text{ X } 100$$

Results of a repeated measures mixed model analysis using GLIMMIX® procedure for PPP, average plantar pressure and contact area were summarized in Table 8, which included all least squares means and standard errors for post-hoc comparison. No interaction between garment and LR was observed for any of the variables. Least squares means and approximate trends are presented in Figure 53 – 56.

Table 8

Plantar Pressure and Contact Area Least Squares Means, Standard Errors, and Significance Levels for Garment and LR Effects

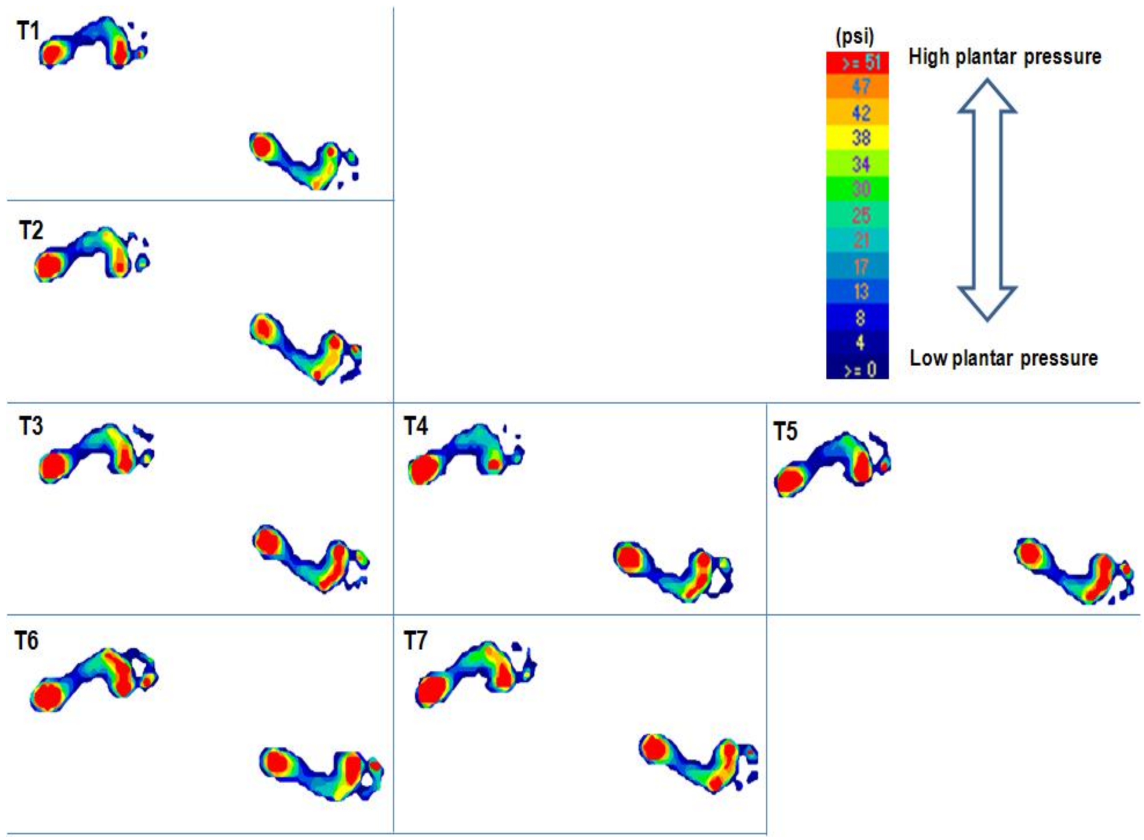
Garment Effect	Garment treatments								
	T1	T2	T3	T4	T5	T6	T7		
Total weight (lb)	1/8	20	40	40	40	60	60		
Weight distribution									
	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	LSMean	Std Error	p-value
***PPP at the forefoot (psi)**	61.55^{c}	65.84^{bc}	66.45^{bc}	68.82^{ab}	70.06^{ab}	71.66^{a}	70.03^{ab}	2.99	**0.0001**
***PPP at the rearfoot (psi)**	59.41^{c}	61.82^{bc}	67.03^{abc}	66.35^{abc}	70.03^{a}	70.28^{a}	68.66^{ab}	4.02	**0.0025**
PPP at the toe (psi)	45.12	45.89	50.87	48.47	49.06	50.91	50.94	6.42	0.1278
***Average plantar pressure (psi)**	24.009^{b}	24.41^{b}	25.88^{ab}	26.11^{ab}	25.57^{ab}	27.46^{a}	26.79^{ab}	1.102	**0.0079**
***Δ Contact area (%)**		0.70^{b}	2.43^{ab}	2.31^{ab}	2.90^{ab}	3.94^{a}	3.82^{a}	0.68	**0.003**

LR Effect	Right		Left		
	LSMean	Std Error	LSMean	Std Error	p-value
***PPP at the toe (psi)**	46.51	6.28	50.996	6.24	**0.0094**

Note. Bold letter with * indicates a statistical significance for the effect.
abc means with a common superscripts in the same row are not significantly different (Tukey, α =0.05).

Peak Plantar Pressure (PPP)

The PPP was observed in the metatarsal heads at the forefoot and ball of the heel as shown in Figure 52. The observed PPPs at the ball of the heel and the metatarsal head are typical plantar pressure patterns to move the body forward for continuous walking.



Note. Red color indicates peak plantar pressure.

Figure 52. Plantar pressure measured by foot sensor mat.

Lord, Reynolds and Hughes (1986) explain the PPP occurs in the ball of the heel, the metatarsal heads and toes sequentially during gait cycle as the foot goes through a sequential walking pattern from heel strike to toe off during the stance phase.

Forefoot

A significant garment effect was found for PPP at the forefoot (p = 0.0001). Post-hoc tests showed a significant difference in PPP (see Table 8). PPPs while wearing T4 (68.82 psi), T5 (70.06 psi), T6 (71.66 psi) and T7 (70.03 psi) were significantly larger than PPP while wearing

T1 (61.55 psi) ($p \leq 0.0027$). PPP while wearing T6 (71.66 psi) was significantly larger than PPP while wearing T2 (65.84 psi) and T3 (66.45 psi) ($p \leq 0.0414$).

A significant linear trend ($p < 0.0001$) across four levels of weight of garment and carrying load was found (see Figure 53); overall, PPP at the forefoot <u>increased</u> with an increase in weight of the garment and carrying loads. There was no significant difference in PPP between the left and the right sides.

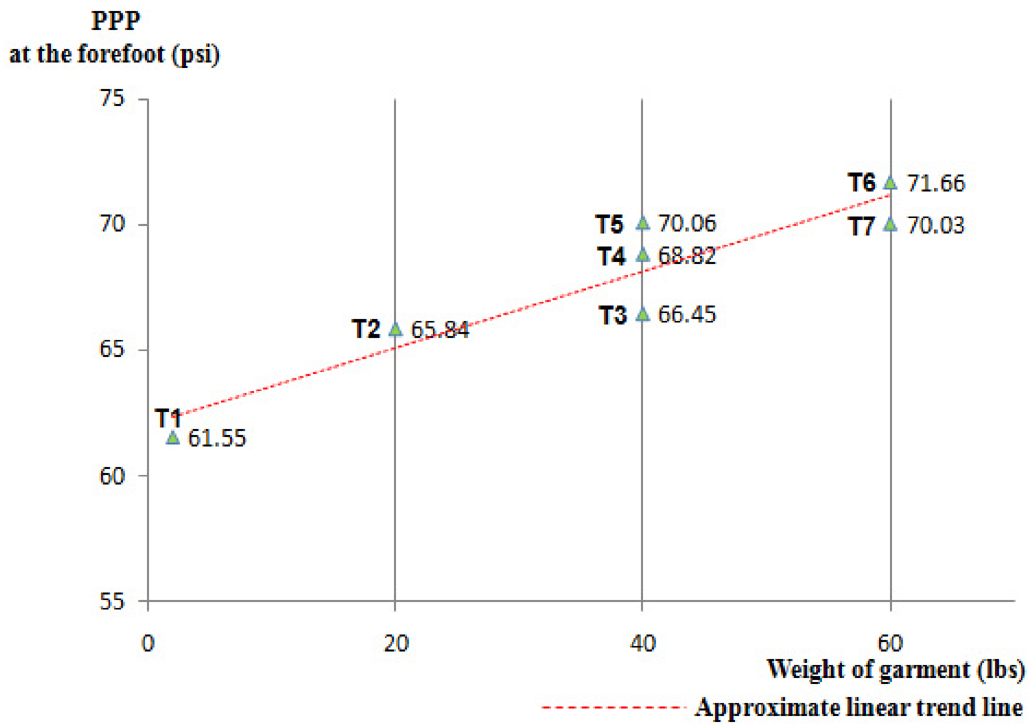


Figure 53. PPP at the forefoot least squares means and trend.

Rearfoot

A significant garment effect was found for PPP at the rearfoot ($p = 0.0025$). Post-hoc tests showed a significant difference in the PPP among the different garment treatments (see Table 8). PPPs while wearing T5 (70.03 psi), T6 (70.28 psi) and T7 (68.66 psi) were significantly larger than PPP while wearing T1 (59.41 psi) ($p \leq 0.0136$). In addition, PPP while wearing T5 (70.03 psi) and T6 (70.28 psi) were significantly larger than PPP while wearing T2 (61.82 psi) ($p \leq 0.0324$). A significant linear trend ($p < 0.0001$) across four levels of weight of garment and carrying load was found as shown in Figure 54. Overall, the PPP at the heel <u>increased</u> with an increase in the weight of garment and carrying loads. There was no significant LR effect on the PPP at the rearfoot. The treatment order was significant ($p = 0.0195$).

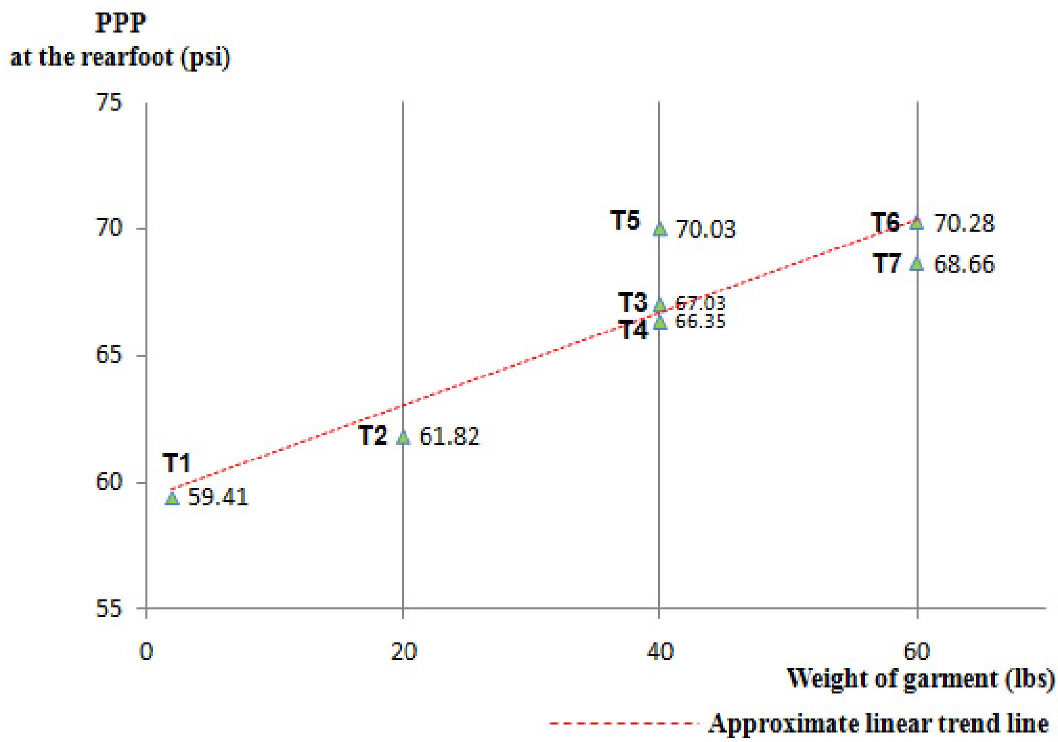


Figure 54. PPP at the rearfoot least squares means and trend.

Toe

No significant garment effect was found for PPP at the toes. PPP on the left foot (50.996 psi) was significantly higher than on the right side (46.51 psi) ($p = 0.0094$).

Average Plantar Pressure

A significant garment effect was found (p = 0.0079). Post-hoc tests showed that the average plantar pressure over the contact area while wearing T6 (27.46 psi) was significantly higher than while wearing T1 (24.009 psi) and T2 (24.41 psi) (see Table 8). A significant linear trend (p = 0.0002) across four levels of weight of garment and carrying load was found as shown in Figure 55. Overall, average plantar pressure <u>increased</u> with an increase in the weight of garment and carrying loads. There was no significant difference in the average plantar pressure between the left and the right sides.

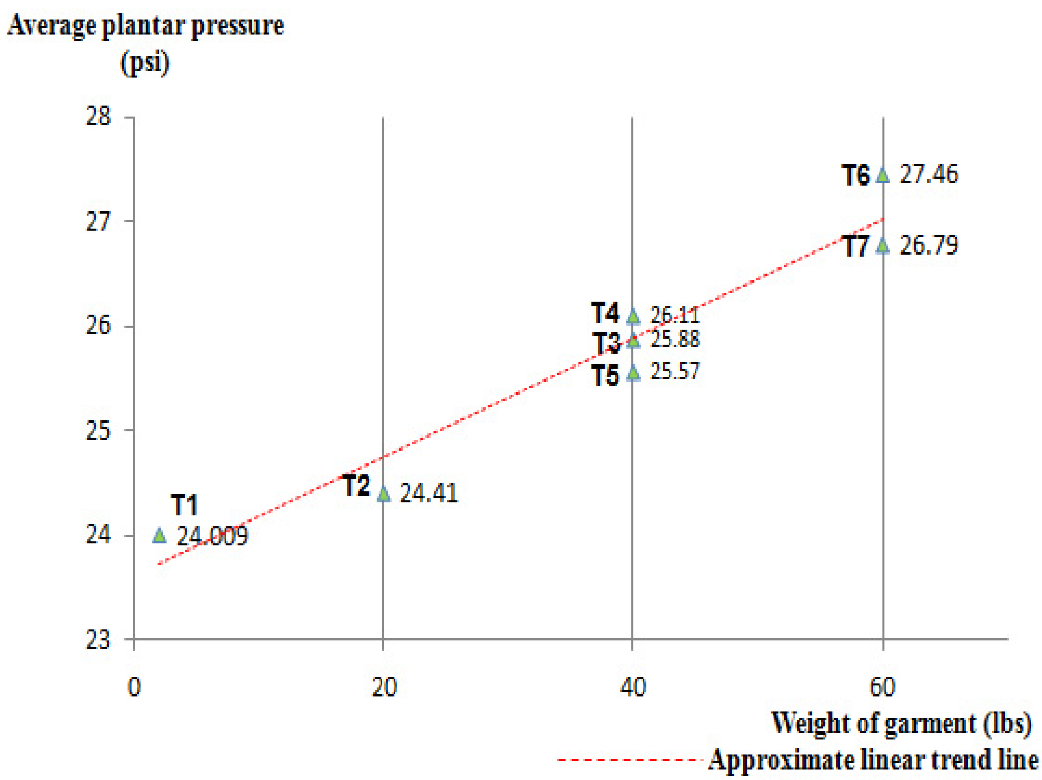


Figure 55. Average plantar pressure least squares means and trend.

Change in Contact Area

A significant garment effect was found (p = 0.003). While wearing T2 through T7, the contact area increased from 0.70 % through 3.94% compared to the contact area found at T1 as shown in Table 8. Post-hoc tests showed that the change in contact area at T6 (3.94%) and T7 (3.82%) were significantly higher than the change at T2 (0.70%) (see Table 8) ($p \leq 0.0073$). A significant linear trend (p = 0.0029) across four levels of weight of garment and carrying load was found as shown in Figure 56. Overall, change in contact area <u>increased</u> with an increase in weight of the garment and carrying loads. There was no significant difference in change in contact area between the left and the right sides.

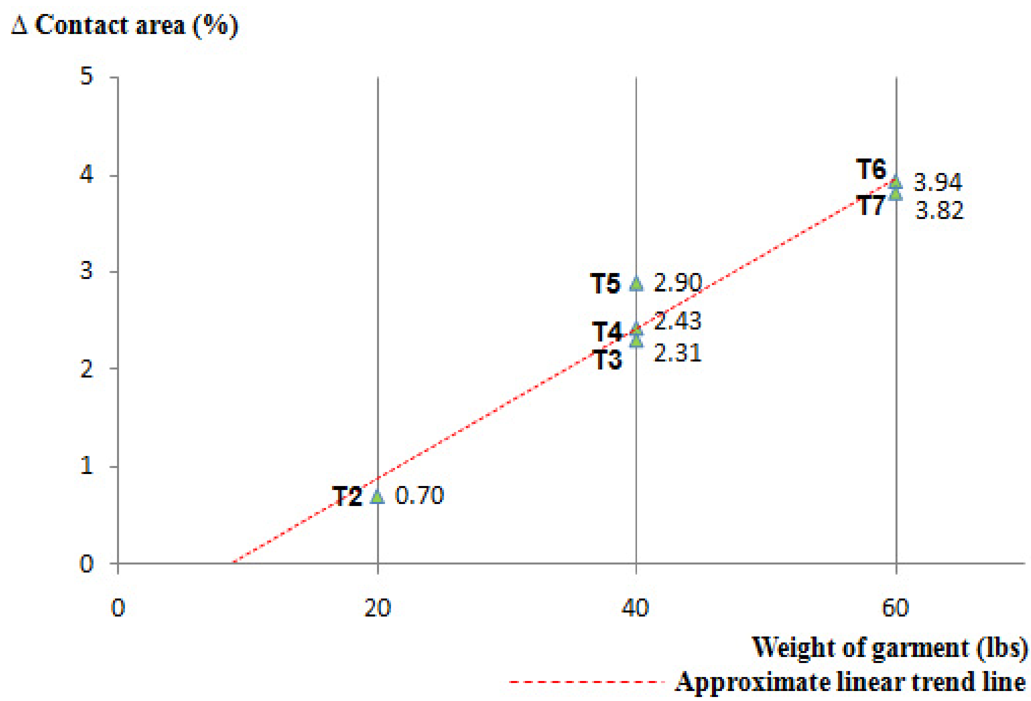


Figure 56. Post-hoc and trend test for the change in contact area.

Summary of Results for Plantar Pressure and Contact Area

In general, an increase in weight of garment and carrying loads significantly increased PPP in the forefoot and rearfoot, average plantar pressure and contact area. When the garment weight was greater than 40 or 60 lbs (T3, T4, T5, T6 and T7), PPP at the forefoot and the rearfoot showed a statistically significant linearly increasing trend. This result implies that the increased peak and average plantar pressure may imply the increased impact force, which results from an increase in magnitude of garment weight and load carriage. This result suggests that the enhanced cushioning is needed in these areas to alleviate impact force and possible foot blisters. There was no significant garment effect on PPP at the toe. Post-hoc tests for PPP showed a significant mean difference only across the weight levels, not within the weight level; in the same weight condition, different weight distribution did not make a difference in PPP at the forefoot and the rearfoot. There was no significant mean difference in PPP between the forefoot and rearfoot ($p = 0.478$).

The significant LR effects showed that subjects had a tendency of experiencing higher PPP at the toe on the left side than on the right side regardless of garment conditions (garment effect was not significant in the PPP at the toe). Average plantar pressure and change in contact area increased to a relatively small extent, compared to PPP at the forefoot and heel. Post-hoc tests for average plantar pressure and change in contact area showed a significant mean difference only across the weight levels, not within the weight level; in the same weight conditions, different weight distribution did not make a difference in the these variables.

Subjective Perceptions about Discomfort and Fatigue

Subjects' perception of ease of movement and perceived discomfort and fatigue were assessed using a prepared ballot with a 5-point Likert type scale (1: No garment effect, 2: very slightly limited movement, 3: slightly limited movement, 4: limited movement and 5: severely limited movement). The ballot was used to assess the ease of movement, after each walking test while wearing each garment treatment. Mean data are given in Table 9. No statistical analysis was performed on these data. Open-ended comments were also permitted and a summary of these are given in the far right-hand column of Table 9.

Overall, as weight of the garment treatment increased, subjects' rating of limitation in movement increased. While wearing T1 (mean = 1), subjects did not feel any limitation in movement or any influence of garment. When subjects wore T2, subjects reported perception of very slight limited movement as shown in Table 9. While wearing the three 40 lb garments (T3, T4 and T5), the mean response rose to 2.93, 2.5 and 2.79 respectively indicating subjects reported that their walking was slightly affected by weight and weight distribution of garment.

While subjects wore one of the garments with an unbalanced weight distribution attached on the front torso (T3 and T4), three subjects reported that their walking seemed to be pulled toward the side where the additional carrying load was placed. Note that while wearing T3, subjects' mean score of perceived impact on walking was greater than mean scores reported for T4 and T5, yet all three loads were 40 lbs. When subjects wore T7, subjects reported the greatest impact of garment treatment on their walking (T7 = 3.93).

In general, subjects' reported mean scores increased with an increase in magnitude of garment and carrying load weight. Subjects reported the greatest garment effect while wearing T3 (40 lb garment with carrying load placed on the non-dominant side of the front torso) and T7 (60 lb garment with carrying load placed on the back).

Discomfort and fatigue were not reported while wearing T1 (1/8 lb). Uncomfortable pressure in the shoulder and heel were first reported while wearing T2 (20 lb OTV). In garment treatments heavier than T2, the number of instances and extent of reported pressure in the shoulder and heel increased with an increase in the weight of the garment and carrying load.

Table 9

Subjective Perceptions about Discomfort and Fatigue

Garment treatments	Weight (lb)	Weight distribution	Mean rating of perceived impact on walking *	Perceived fatigue and discomfort
T1	1/8		1	- No fatigue and discomfort
T2	20		1.5	- Shoulder: mild pressure on both sides (S1, S2) - Heel: Pressure on both sides (S2)
T3	40		2.93	- Neck: uncomfortable pressure (S1, S2, S7) more strain on the right (S7) - Shoulder: uncomfortable pressure and fatigue (S2, S6, S7) more pressure and strain on the left (S2, S6) - Abdomen: strain on the left to stand straight due to unbalanced weight attached to the left (S5) - Back: strain in the lower back (S1) - Heel: uncomfortable pressure on both sides (S2) - Body balance: feeling of being pulled toward the left while walking and feeling of unbalance (S5, S7)
T4	40		2.5	- Neck: Uncomfortable pressure on both sides (S7) - Shoulder: uncomfortable pressure (S1, S2) mild fatigue (S1) feels heavy (S2) - Abdomen: feel heavier on the right due to the attached weight (S6) - Back: strain int the lower back (S1) - Heel: uncomfortable pressure on both sides (S2) - Body balance: feels the garment pulls down on the right side, which makes me walk on the side where the weight was placed (S5) weight on the right changed my walking slightly (S6)
T5	40		2.79	- Neck: uncomfortable pressure (S1, S6, S7) - Shoulder: uncomfortable pressure (S1, S2, S3, S7) - Back: strain in the lower back (S2, S4) fatigue in the upper and lower back (S4) - Heel: a lot of uncomfortable pressure on both side (S2, S6)
T6	60		2.86	- Neck: uncomfortable pressure (S1, S2, S6) - Shoulder: uncomfortable pressure (S1, S2, S3, S5, S6, S7) feel most pressure is on the shoulder (S6) greater fatigue (S1) - Back: strain (S1, S2, S5, S7) - Heel: a lot of pressure (S2, S6)
T7	60		3.93	- Neck: uncomfortable pressure (S1, S6) excessive pressure on the back neck (S4) - Shoulder: very uncomfortable and stressful pressure (S1, S2, S3, S5, S6, S7) sore in the shoulder (S5) fatigue (S7) - Back: painful (S1), strain in the lower back (S2, S5) - Knee: strenuous (S6) - Heel: a lot of pressure (S2, S6) - Other comments: chest and neck get pulled back due to weight on the back (S3) feels shoulder pulled back and down (S7)

Note. * Averaged response from participants.
Likert scale response indicates 1: No effect, 2: Very slightly limited movement, 3: Slightly limited movement, 4: Limited movement, 5: Severely limited movement

While subjects wore the three 40 lb garments (T3, T4 and T5), subjects felt uncomfortable pressure in the neck and strain in the lower back, in addition to feeling uncomfortable pressure in the shoulder and heel. In particular, when subjects wore T3 and T4 with asymmetric weight distribution on the front torso, subjects reported greater pressure and strain on one side than the other side. Subjects' responses were as follows: "greater strain on the right neck" (T3), "greater strain on the left abdomen to stand straight due to unbalanced weight placed on the left" (T3), "more pressure and strain on the left shoulder" (T3), "feel heavier on the right side" (T4). Shoulder fatigue was also reported both in T3 and T4 conditions. When wearing T5 with symmetric weight distribution, unbalanced walking and asymmetric strain and discomfort were not reported. Subjects' mean ratings about the limited movement caused by garment treatments was higher while wearing T3 (2.93) than while wearing T4 (2.5) or T5 (2.79). Therefore, subjects' reported limitations in walking were greater in T3 than in T4 and T5.

Subjects reported greater pressure in the neck and shoulder, lower back and heel while wearing the 60 lb garments (T6 and T7), than while wearing the 40 lb garments (T3, T4 and T5). In particular, when wearing T7 subjects reported the greatest pressure on the back neck, and soreness in the shoulder and low back and strenuous movement in the knee.

Summary of All Results

This study investigated multiple dependent variables to identify the impact of weight and weight distribution of ballistic vest and carrying load on lower body movement while walking. Results of all quantitative data analyses are summarized in Table 10.

A significant garment effect was found for four temporal and distance parameters (stance phase, swing phase, double support and stride length) as shown in Table 10.

Table 10

Summary of Significant Results

	Significant garment effect		Significant LR effect	
Temporal & distance parameters	**Variables (4/6)**	**Significant linear trend**	**Variables (2/6)**	**Post-hoc comparison**
	Stance phase	Increase with weight level	Stance phase	Right > Left
	Swing phase	Decrease with weight level	Swing phase	Right < Left
	Double support	Increase with weight level		
	Stride length	Decrease with weight level		
ROM	**Variables (6/10)**	**Significant linear trend**	**Variables (4/10)**	**Post-hoc comparison**
	Pelvic obliquity	Decrease with weight level	Pelvic tilt	Right < Left
	Hip adduction-abduction		Pelvic rotation	Right > Left
	Pelvic tilt	Increase with weight level	Hip rotation	Right < Left
	Hip flexion-extension	Increase with weight level	Knee rotation	Right > Left
	Pelvic rotation	Decrease with weight level		
	Hip rotation			
Joint angle$_{max}$	**Variables (2/19)**	**Significant linear trend**	**Variables (6/19)**	**Post-hoc comparison**
	Pelvic intrarotation	Decrease with weight level	Pelvic obliquity - upward	Right > Left
	Pelvic extrarotation	Decrease with weight level	Pelvic obliquity - downward	Right < Left
			Hip adduction	Right > Left
			Pelvic tilt - anterior	Right < Left
			Hip flexion	Right < Left
			Ankle extrarotation	Right > Left
EMG	**Variables (1/4)**	**Significant linear trend**	**Variables (2/4)**	**Post-hoc comparison**
	Rectus femoris	Increase with weight level	Bicep femoris long head	Right < Left
			Medial gastrocnemius	Right < Left
Plantar pressure & contact area	**Variables (4/5)**	**Significant linear trend**	**Variables (1/5)**	**Post-hoc comparison**
	PPP at the forefoot	Increase with weight level	PPP at the toe	Right < Left
	PPP at the rearfoot	Increase with weight level		
	Average plantar pressure	Increase with weight level		
	Change in contact area	Increase with weight level		

Note. The first number inside the parenthesis indicates the number of variables found statistically significant result.
The second number inside the parenthesis indicates the number of total variables analyzed.

Significant garment effect on temporal and distance parameters of walking indicated that an increase in garment weight significantly increased the time that the foot was in contact with the floor to maintain body balance under weight-bearing conditions. This result was supported by a significant garment effect on increased stance phase and double support, as well as a decreased swing time. The stance phase and double support linearly increased with an increase in the garment and carrying load weight, and the swing phase linearly decreased with an increase in the garment and carrying load weight. A significant difference by weight distribution was found for the temporal and distance parameters. T3 caused the longest stance phase and double support as well as the shortest swing time. This result suggests that the weight placed on the left torso (non-dominant side) may create the greatest instability while walking, which causes the body to increase the foot contact time with the floor to maintain body balance.

Significant LR effects were found for stance phase and swing phase, which indicated that the right foot contacted the floor for a longer time than the left foot did. This result implies that the right handed subjects had a tendency to use the right foot more dominantly for weight bearing and maintaining body balance.

A significant garment effect was found for six ROMs (pelvic obliquity, hip adduction-abduction, pelvic tilt, hip flexion-extension, pelvic rotation and hip rotation) and for two joint angle_{max} (pelvic intrarotation and extrarotation) as shown in Table 10. These results showed that the weight and weight distribution of each garment changed joint movement in the pelvis and hip. An increase in garment weight increased pelvic tilt. In particular, T7 showed the greatest increase in pelvic tilt which appears to result from the placement of the carrying load on the back. The additional weight on the back may create the greatest forward lean of the trunk. Post-hoc tests showed that weight distribution makes a difference in pelvic tilt; pelvic tilt while wearing T7 was significantly greater than while wearing T6 despite the same weight level. On the other hand, T6 with a balanced weight distribution showed no significantly different pelvic tilt as compared to the lighter weight garment treatments (T1 through T5). An increase in garment weight significantly decreased pelvic rotation and obliquity by adding a physical burden to the lower body.

ROM for hip adduction-abduction and hip rotation gradually decreased when subjects wore the 20 lb (T2) and 40 lb garments (T3, T4, T5), but increased when subjects wore the 60 lb garments (T6 and T7). ROM for hip flexion-extension showed an overall linearly increasing trend with an increase in weight of the garment and carrying load with an exception of the 40 lb garment treatments (T3, T4 and T5).

A significant LR effect was found for multiple movements at the pelvis, hip, knee and ankle as shown in Table 10. However, there was no significant interaction between garment effect and LR effect. Therefore, this result indicates that the right handed subjects in this study tended to use a certain movement pattern, which resulted in using one side more than the other for knee rotation, pelvic tilt and hip rotation regardless of garment treatment.

A significant garment effect was found for rectus femoris during the gait cycle. The peak amplitude of EMG for rectus femoris linearly increased as weight of the garment and carrying load increased. The two peak curves revealed noticeable mean differences occurred about 10% (loading phase) and 30-60% (terminal stance and pre-swing) of the gait cycle, where the body’s center of gravity moves from one leg to the other. This result implies that the increased garment weight increases the muscle activity of the rectus femoris to maintain body balance and stability under weight-bearing conditions. Post-hoc tests further identified significant mean differences between (T1 and T7) and (T3 and T7). This result implies that T7 with the 40 lb carrying load placed on the back may require the greatest muscle force in the rectus femoris to maintain stable walking.

A significant garment effect was found for four variables measuring plantar pressure and contact area. These results showed a significant increase in peak plantar pressure at the forefoot and the rearfoot, average plantar pressure and contact area with an increase in magnitude of the weight of garment and carrying loads. Significant mean difference of PPP, average plantar pressure and contact area were found only between the weight levels, not within each level, which implies weight distribution in the same weight level does not affect the peak plantar pressure, average plantar pressure and contact area, in this study. A significant LR effect was found for PPP at the toe as shown in Table 10.

Subjects' rating of perceived limitation in movement while walking, as assessed a 5-point ballot, increased as the weight of the garment and carrying load increased. In particular, when subjects wore T3 and T7, subjects reported the greatest limitation in their walking. Subjects' perception about discomfort and fatigue indicated that wearing the OTV with a carrying load can cause uncomfortable pressure, fatigue and muscle pain in the neck, shoulder, and low back. In addition, when subjects wore T3 and T4 with asymmetric weight on the front torso, subjects felt their walking became unbalanced due to the weight. Furthermore, subjects felt strain in the neck and abdomen more on one side than the other. Subjects reported the greatest uncomfortable pressure, soreness, and fatigue in the neck, shoulder, lower back, and even the knee when they wore T7.

CHAPTER V

SUMMARY AND CONCLUSIONS

This study investigated the impact of the weight and weight distribution of the OTV and additional carrying load attached to the OTV on lower body movement by comparing walking patterns, joint angle$_{max}$, ROM, EMG on leg muscles, plantar pressure, pressure contact area and perception about discomfort and fatigue during walking. Gait analysis, which has been used for clinical applications to diagnose and rehabilitate orthopedic problems (Davis, Ounpuu, Tyburski & Gage, 1991), was used to identify biomechanical changes of the lower limbs and changes in walking patterns while wearing different garment treatments with varying weight and weight distribution.

A human subject test was conducted to test hypotheses made based on research objectives. Seven healthy male ROTC students participated in this study. The independent variable was garment condition with seven levels. Treatment 1 was a pair of snuggly fitting sports shorts (1/8 lb). Treatment 2 included additionally wearing a 20 lb OTV with front and back ceramic plates. The OTV consisted of Cordura ® outer and inner shells, soft armor panel inserts made of multiple layers of Kevlar, and two ceramic plates.

The OTV has webbing on the front and back called a ‘Molle system’, which is designed for load carriage.

Treatments 3, 4 and 5 had a 20 lb carrying load attached to the front of the OTV. In treatment 3, a 20 lb carrying load was attached to the left front of the OTV, while in treatment 4, the 20 lb carrying load was attached to the right front of the OTV. Treatment 5 also had the same 20 lb, but the weight was evenly distributed to the left (10 lb) and right (10 lb) front sides. Treatments 6 and 7 had a 40 lb carrying load attached to the OTV. Treatment 6 had four pouches containing 10 lb each. Two pouches were attached to the front (left and right sides) and two pouches were attached to the back (left and right sides). Treatment 7 had two pouches containing 20 lb each. These two pouches were attached to the back (left and right sides). Therefore, this garment treatment included four levels of weight (1/8 lb, 20 lb, 40 lb and 60 lb) with different weight distributions.

Dependent variables characterizing the lower body movement were measured while subjects walked wearing the treatment garments. A human subject test included measurement of walking patterns, joint movement, EMG, plantar pressure and pressure contact area. In the first walking test using motion capture and EMG system, temporal and distance parameters (stance phase, swing phase, double support, stride length, step width and walking speed), 19 joint $angle_{max}$, 10 ROMs and the normalized amplitude of EMG signals on four leg muscles (rectus femoris, bicep femoris, tibialis anterior, and medial gastrocnemius), were measured as dependent variables. This walking test was repeated five times per garment condition. In the second walking test using a foot pressure sensor mat, peak plantar pressure, average plantar pressure and the change in contact area of the foot were measured as dependent variables. This test was repeated four times per garment treatment. To avoid a possible effect of the previous test on the subsequent test, the order of measurements were conducted using a crossover design (Latin Square Design) with repeated measures. Data were statistically analyzed through a mixed models repeated measures analyses using either the SAS/MIXED® procedure or SAS/GLIMMIX®

procedure. When garment effect was significant, post-hoc tests of Tukey pairwise comparisons and trend analyses were further conducted.

Two to five minutes of rest time were given between the garment treatments. During the rest time after performing the walking test for each garment condition, the researcher asked the subjects' perception about discomfort, fatigue and perceived impact of garment on their walking. Subjects' ratings about limited movement were averaged and their comments about perceived discomfort and fatigue were summarized.

This study confirmed that weight and weight distribution of the OTV and carrying load placed on the upper body significantly changed walking patterns, joint movements, EMG on the leg muscles and plantar pressure and contact area at the foot.

This study found possible negative impacts of garment weight on soldiers' lower body mobility. As the weight of the garment and carrying load increased, the foot contacted the floor for a longer period of time as supported by an increased stance time and double support, as well as decreased swing time as shown in Table 4 on page 54. These statistically significant results, which are likely due to the body's effort to maintain balance and stability under weight-bearing conditions, concur with Kinoshita's (1985) backpack study. Kinoshita explained that an increase in load magnitude typically shortens swing time, which leads to a shorter stride length. Kinoshita (1985) attributed this change to the human body's adaptive response to allow for transferring body weight from one leg to the other under weight-bearing conditions. However, the shortened stride length found in this study could have a negative impact on soldiers' mobility because it requires more rapid and frequent strides to maintain the constant speed, which may increase overall energy expenditure with resulting fatigue.

An increase in weight of garment and carrying load also increased pelvic tilt, which is closely related to forward lean of the trunk. Forward lean of the trunk is helpful to minimize the energy expenditure of carrying load and to stabilize the body's center of mass (Birrell and Haslam, 2009). However, the forward inclination of the trunk and its resultant pelvic tilt resulting from weight bearing has been reported to possibly lead to chronic lumbar pain and disorder (Smith et al., 2006; Birrell and Haslam, 2009).

Decreased pelvic rotation resulting from an increase in weight may have a negative impact on soldiers' mobility. Reinhardt (n.d.) claimed that limited rotation of the pelvis decreases the efficiency of walking and running by limiting the leg's swing force to move the body forward, which also results in a shorter stride length. Therefore, increasing stride frequency is necessary to maintain the walking speed under the weight bearing condition (Kinoshita, 1985; Birrell and Haslam, 2009). However, a decrease in pelvic rotation due to weight-bearing conditions seems to be the human body's natural protective adaptation. Birrell and Haslam (2009) claimed that the decreased pelvic rotation while wearing a backpack is a human body's natural reaction to minimize the production of torque (the application of a force at point of rotation or at a perpendicular distance to a joint) in the torso as an attempt to attenuate the risk of low back pain and possible injuries. LaFiandra, Wagenaar, Holt and Obusek (2003) also claimed that an increase in upper body torque during load carriage can lead to low back injury. Smith et al. (2006) in their backpack study also found decreased pelvic rotation and they claimed that the observed limited pelvic rotation is probably for maintaining dynamic stability while walking. Insufficient pelvic rotation resulting from garment weight and carrying load appears to lead to a shorter stride length as found in this study. Birrell and Haslam (2009 & 2010), in their studies of military load carriage showed that a weight bearing condition limits pelvic rotation, which leads to shorter stride length and wider step width by restricting the leg's swing to propel the body forward and each foot's ability to return to its normal position.

In general, ROM for hip flexion-extension increased as garment weight increased. ROM for hip adduction- abduction and hip rotation showed a decreasing trend up to the 40 lb weight level. However, at the 60 lb weight level in T6 and T7, ROM for adduction- abduction and hip rotation increased. This result appears to associate the compensational relationship between the pelvis and hip movement as suggested by LaFiandra et al. (2003). Their study found increased hip excursion compensating for decreased pelvic rotation while subjects wore backpacks that weighed 40% of their body mass. In the current study, the magnitude of weight linearly decreased ROM for pelvic rotation. Similarly, in this study, ROM for hip rotation decreased with an increase in the weight of the garment up to the 40lb condition. However, with the 60 lb load condition, ROM for hip rotation increased, probably due to the need for compensation for the decreased pelvic rotation from such a heavy load as shown in a previous study by LaFiandra et al. (2003). Birrel and Haslam's (2009) study also showed a similar data pattern for hip rotation with a decreasing trend (between unload conditions and 8 kg of carrying load) and an increasing trend (between 8 kg , 16kg, 24 kg and 32kg) at controlled walking speed (1.5 m/s).

The increased peak amplitude of EMG found for rectus femoris implies weight bearing conditions could cause early muscle fatigue in the muscle by repeatedly applying increased physical force. Ricamato and Hidler (2005) showed that high amplitude of EMG can lead to early fatigue by increasing metabolic energy consumption and also cause degenerative joint problems in the long term. Such negative impacts can be greater at faster walking and running speeds. Ricamato and Hidler (2005) showed that an increase in walking speed increased the amplitude of EMG in leg muscles. Byrne, O'Keeffe, Donnelly and Lyons (2007) explained that an increase in walking speed requires larger accelerative and deccelerative muscle force resulting in a greater arc of motion of the leg segment. In the current study, a significant garment effect was found for only rectus femoris, which may be because of the slow walking speed used for this study. Shiavi, Bugle and Limbird (1987) and Milner, Basmajian and Quanbury (1971) showed that greatly varied and individualized EMG patterns in leg muscles were observed at slower walking speeds, while more identical and homogeneous EMG patterns among muscles and subjects were observed at faster walking speeds. They attributed this finding to dominant muscle working mechanism at faster walking speeds. A significant LR effect was found for bicep femoris long head and medial gastrocnemius as shown in Table 10. However, there was no significant interaction between garment effect and LR effect. Therefore, this result indicates that the right handed subjects in this study had a certain movement pattern that applied more physical force on one side than the other side for walking regardless of the garment conditions.

Increased peak plantar pressure (PPP) in the forefoot and rearfoot suggest the potential for foot blisters and overuse injuries at the joints resulting from excessive impact forces that soldiers can experience in weight bearing working conditions. Increased PPP implies an increase in the impact force with an increase in magnitude of weight of the garment and carrying load. Increased impact force has been reported to be a major risk of metatarsal stress fractures, knee joint problems and other overuse injuries by previous load carriage studies (Cavanagh & Lafortune, 1980; Knapik, 2001). In the current study, the increased PPP was observed around the metatarsal heads and heel, which suggests enhanced cushioning is needed in these areas to alleviate possible foot blisters and musculoskeletal injuries. It should be recalled, however, subjects in this study walked barefoot and the author has no direct knowledge of the military boots in use today.

There was no significant mean difference in the observed PPP between the forefoot and the rearfoot. However, the literature shows that the forefoot is more vulnerable to possible foot blisters and ulcers caused by repeated loading of high peak pressure not only because the forefoot has soft skin tissue while the rear foot has thick sturdy tissue (Zou, Mueller & Lott, 2007), but also because the forefoot sustains the body weight for a longer time during a walking cycle (Lord et al., 1986). Lord et al. (1986) found weight bearing time of the forefoot was three times as long as that of the rearfoot.

The increased contact area found in this study with an increase in garment weight indicates that subjects' feet flattened and their arches were lowered. In sum, the increased peak plantar pressure and contact area found with an increase in weight of garment and carrying load implies the potential for foot blisters and overuse injuries resulting from excessive impact forces. In particular, the literature shows that foot blisters can negatively affect soldiers' mobility and even their health. Yavuz and Davis (2010) claimed foot blisters are common military injuries which compromise the effectiveness of military operation and training. Patterson, Woolley and Lednar (1994) reported that 42% of the cadets in a ROTC (Reserve Officers' Training Corps) experienced foot blisters at summer military training. Polliack and Scheinberg (2006) reported that foot blisters can lead to pains, uncomfortable movement, cellulitis and even toxic shock syndrome.

This study also showed that negative impacts of an increase in garment weight can be influenced by weight distribution of the carrying load. Treatments T3, T4 and T5 had a 20 lb carrying load attached to the front of the OTV. T3 load was positioned on the left front and T4 load was positioned on the right front. T5 had a 10 lb load positioned on the right front and a 10 lb on the left front. When subjects wore T3, they experienced the longest foot contact time as shown by the greatest stance phase and double support and smallest swing time. It is interesting that the asymmetrically placed weight on the left side (non-dominant side) caused these results for the right-handed subjects in this study. The increased foot contact time may have occurred to achieve body balance and stability. This finding has practical implications for soldiers. Heavier loads should not be placed on the non-dominant side of the front of the body. This may be counterintuitive, since reaching for an object with the dominant hand from the non-dominant side seems natural and easier.

Treatments T6 and T7 had a 40 lb load attached to the OTV. T6 had two 10 lb loads positioned on the right and left front OTV and two 10 lb loads positioned on the right and left back OTV. T7 had two 20 lb loads positioned on the right and left back OTV. The greatest pelvic tilt was found when subjects wore T7. Weight placed on the back in the T7 condition increased the forward lean of the torso, which significantly increased pelvic tilt, which has been reported to be a cause of chronic lumbar pain (Smith et al., 2006). Pelvic tilt while wearing T6 was not significantly different from pelvic tilt while wearing lighter weight garment treatments (T1 through T5). These results suggest that balanced (right, left, front and back) weight distribution placed on the torso may be helpful to avoid forward lean of the trunk, which may alleviate the impact of weight on the pelvic tilt in the sagittal plane. The issue of weight distribution although previously investigated, appears to warrant further research especially regarding placement of the load on the dominant and non-dominant sides and front and back.

Results of EMG measurement confirmed a significantly higher amplitude for rectus femoris while wearing T7 (carrying load was placed on the back), which may cause early muscle fatigue. Cook and Neumann (1987), in their backpack study, also showed that unbalanced weight bearing can cause early muscle fatigue by increasing muscular activity on the side opposite the load and greater compressive forces down on the spine.

Subjects also reported higher mean scores while wearing T3 and T7 indicating more limitations in walking. In addition, subjects reported unnatural walking while wearing T3 and T4 which had unbalanced carrying loads placed on the right and left front torso. When subjects primarily wore T 3 through T7, they reported comments about uncomfortable pressure, muscle strain and pain in the neck, shoulder, lower back and heels. Perhaps, had the weight bearing not been placed on the shoulders only, these negative comments may not have occurred. In sum, results of quantitative data (temporal and distance parameters, ROM, joint angle$_{max}$ and EMG) and subjects' perception about ease of movement, discomfort and fatigue demonstrated that unbalanced weight distributions such as T3 and T4 may have negatively impact the lower body movement while walking.

This study demonstrated the impact of weight and weight distribution of a body armor vest and load carriage on soldiers' lower body movement. The results suggest that a body armor vest should ideally weigh less than the OTV and include design features that reduce the heavy dependence of weight bearing on the shoulders. An improved vest weight suspension system such as a hip belt / or shoulder load lifter, which help distribute weight between the shoulders and the pelvis may alleviate or reduce some of the negative impact found in this study. The improved OTV (IOTV) fielded for the US Army in 2007 includes a hip belt designed to distribute vest weight between the shoulders and lower torso (*Army magazine,* 2010). A vest that incorporates a carriage system should contain design features to ensure balanced (left, right, front and back) weight distribution to improve soldiers' mobility with less impact on their joints and muscles. Training in the use of such a vest should include an emphasis on the importance of weight distribution of the load.

In addition, using padded insoles made of a high level of cushioning in the forefoot (especially in the metatarsal heads where most weight-bearing is concentrated) and heel is suggested to minimize the negative effects of impact forces including foot blisters. Wearing socks made of low friction materials may also be effective in minimizing the possibility of foot blisters by allowing more sliding between the plantar foot and the socks. Wearing a proper size military boot is also important because it can minimize unnecessary rolling of the foot inside the boots, which increases friction between the plantar foot, socks and boots.

Recommendations for Future Research

1. This study used a crossover design (Latin Square Design) with repeated measures to control the impact of order of garment treatments. Furthermore, a rest time was specified after each garment treatment test was completed in an effort to control a possible order effect of garment treatments.

Although this experimental design successfully blocked the impact of order of garment treatment in almost all dependent variables as proved by insignificant effect of treatment order, significant order effects were found for four dependent variables (ROM for ankle rotation, joint angle$_{max}$ for hip flexion and hip extrarotation, and PPP at the rearfoot). For a future study, an experimental design that includes balancing for residual effects of treatment, recruiting at least two subjects for each sequence of treatments, and longer rest time among treatments are recommended.

2. This study conducted a human subject test with the task of walking with a self-preferred walking speed. A future study could test the same variables at different walking speeds. The experiments could focus on the relationships between walking speed, ROM, EMG and walking patterns to allow comparison with previous studies that show different results at different walking speeds.

3. A future study could test the impact of actual military load carriage systems and personal items required for training and combat situation while walking or running a longer distance in outdoor conditions, which may provide practical implications.

4. This study measured EMG of only four major muscles. Exploring more muscles may provide a more comprehensive understanding of the change in muscle movement by garment conditions.

5. This study focused on measuring the impact of weight and weight distribution placed on the upper body on lower body movement especially walking. A future study could explore other active movements in the lower body and upper body.

6. A future study could include a variety of body types with different body mass, height and other physical characteristics expected to influence results.

7. A future study could explore female soldiers' movements under different garment conditions with varying weight and weight distribution.

8. A future study could determine the impact of wearing the IOTV (Improved Outer Tactical Vest) on lower body movements. A comparison with the present study results could be done to demonstrate whether the design of the IOTV results in an improvement in variables assessing lower body movement.

9. A future study could investigate weight distribution including load carriage on the front and back, left and right sides of the torso at various weight levels.

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APPPENDICES

Appendix A

Approved IRB (Institutional Review Board)

Oklahoma State University Institutional Review Board

Date	Tuesday, January 11, 2011	Protocol Expires:	10/14/2011
IRB Application	HE1069		
Proposal Title:	Impact of Body Armor and Load Carriage		
Reviewed and Processed as:	Expedited **Modification**		

Status Recommended by Reviewer(s) **Approved**

Principal Investigator(s) :

Huiju Park	Donna Branson
412 HES	431 HES
Stillwater, OK 74078	Stillwater, OK 74078

The requested modification to this IRB protocol has been approved. Please note that the original expiration date of the protocol has not changed. The IRB office MUST be notified in writing when a project is complete. All approved projects are subject to monitoring by the IRB

X The final versions of any printed recruitment, consent and assent documents bearing the IRB approval stamp are attached to this letter. These are the versions that must be used during the study.

Signature : Shelia M. Kennison

Shelia Kennison, Chair, OSU Institutional Review Board

Tuesday, January 11, 2011
Date

Appendix B

Advertisement Flyer for Subject Recruiting

WE NEED YOUR HELP

- Be part of our research for improvement in soldier mobility

- Are you a male ROTC?
- Are you over 18 year old?
- Are you healthy (no orthopedic medical history)?
- Are you between 5.9 and 6.1 feet tall & between 155 and 180pounds ?
- Are you right handed?

!!! If you say yes, you may be eligible to participate in our research!!!

- **Purpose of Research:**
  To understand the impact of wearing ballistic vest and carrying load on lower body movement during walking
- **Participants will wear seven types of garments with markers and electromyography sensors attached to the skin and walk about five meters at normal speed and movement will be recorded and analyzed by a motion capture system**

- **$ 70 will be given to participants who complete all required tests**

- **This study is being conducted at the IPART Laboratories at Venture I in the Oklahoma Technology and Research Park located at 1110 S. Innovation Way, Stillwater**

- **If interested, please contact the researcher, Huiju Park by email (hui.park@okstate.edu) or phone (405-612-2979) or advisor Dr. Donna Branson (donna.branson@okstate.edu).**

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Appendix C

Script for Subject Recruiting

Thank you for giving me an opportunity to introduce my research to you.

My name is Huiju Park and I am a Ph.D. student studying apparel design in the Dept. of Design, Housing and Merchandising. Currently I am conducting an experiment for my dissertation entitled ‘Impact of Body Armor and Load Carriage on Lower Body Movement’. This study will investigate the effect of weight of armor and weight distribution of load carriage attached to an outer tactical vest (OTV) on lower body movement while walking by using motion capture technology.

I am looking for volunteer male participants without any orthopedic problems, who have experience wearing the OTV. Subjects must be between 5.9 and 6.3 feet tall, weigh between 155 and 230 pounds and be a right-handed. If a volunteer meets these requirements, the volunteer will don a test ballistic vest so that the researcher can check the fit of the vest. If all conditions are met, the volunteer can participate in the human subject test.

The test will be conducted at the IPART Laboratory at Venture I at the Oklahoma Technology and Research Park located at 1110 S. Innovation Way, Stillwater. Participants will wear seven types of garments with markers and electromyography sensors attached to their skin and walk about five meters at normal speed. The weight of the garments range between 20 lbs and 60 lbs. Movement will be recorded and analyzed by a motion capture system. The test will be about 3.5 to 4 hours long. Seventy dollars will be given to participants who complete all required tests.

Data will only be used for the research purpose and there are no greater risks than risk in the normal life in this study. Only the researchers will see the data, so your personal information will be confidential.

If you have an interest in participating in this study, raise your hand now. If you have any questions about this study and experimental procedures, feel free to contact me by email or cell phone given on my business card. Thank you for your time.

Appendix D

Medical History Check List

Name: ___________________________ Date: ___________________

Please answer the following questions.

Have you had any injuries and/or surgeries for the areas of the body listed below?

If yes, please explain.

- Neck: Injury ____yes ____no / Surgery ____yes ____no / Explain: ____________________
- Shoulder: Injury ____yes ____no / Surgery ____yes ____no / Explain: _________________
- Back: Injury ____yes ____no / Surgery ____yes ____no / Explain: _____________________
- Waist: Injury ____yes ____no / Surgery ____yes ____no / Explain: _____________________
- Hip: Injury ____yes ____no / Surgery ____yes ____no / Explain: ______________________
- Knee: Injury ____yes ____no / Surgery ____yes ____no / Explain: _____________________
- Ankle: Injury ____yes ____no / Surgery ____yes ____no / Explain: _____________________
- Thigh: Injury ____yes ____no / Surgery ____yes ____no / Explain: ____________________
- Calf: Injury ____yes ____no / Surgery ____yes ____no / Explain: ______________________
- Foot: Injury ____yes ____no / Surgery ____yes ____no / Explain: ______________________

Appendix E

Informed Consent Form

Approved 1/11/11
Expires 10/14/11
IRB # HE-10-69

CONSENT TO PARTICIPATE IN A RESEARCH STUDY
OKLAHOMA STATE UNIVERSITY

PROJECT TITLE: Impact of Body Armor and Load Carriage on Lower Body Movement

INVESTIGATORS: Huiju Park (M. S.), and Dr. Donna Branson, Oklahoma State University

PURPOSE:
The purpose of this research is to explore the impact of wearing a ballistic vest and carrying selected loads on lower body movement during walking and suggest mobility measurement methodology based on motion capture and electromyography (EMG) technology.

PROCEDURES:
Pre-experiment procedure (10 minutes)
The project will involve seven sequential walking tests while wearing seven different garment treatments. Before testing, you will be asked to wear the control garment (snuggly fitting sports shorts) provided by a researcher. The anthropometric measurements of your lower body will be conducted on a medical bed. Pelvic length, pelvic width, knee diameter, ankle diameter and leg length will be measured.
To place EMG probes and electrode on your leg muscles, the researcher will let you know of eight locations where EMG sensors will be placed on your legs. You will be asked to shave the area of skin, where the EMG probes will be placed, with a razor provided by the researcher in order to minimize possible noise of signal caused by body hair. Then you will rub the shaved skin area with ECG skin preparation gel (Nuprep®, Weaver and Company) to lower skin impedance which may cause incorrect data collection. Then you will completely wipe the gel with soft paper towel and clean the shaved skin area with an alcohol pad provided by the researcher. You will wait for the alcohol to completely evaporate. During this skin preparation process, temporary mild skin irritation is possible.
Then, eight (four right and four left) EMG probes will be snapped onto EMG electrode which will adhere to the skin due to the adhesive surface of the electrodes. EMG probes will detect electronic signals of your muscle activity.

Experiment
1) MVIC (Maximum voluntary isometric contraction) Test (15 minutes)
The researcher will help you take a specific posture for each muscle's MVIC test. You will be asked to maintain the instructed posture for five seconds by resisting the force which the researcher manually applies toward a specific direction to try to change your leg's posture in the MVIC conditions. After completing each MVIC test, you will be given 30 seconds to rest. Each MVIC test will be conducted three times for each leg muscle. A total of 15 minutes will be taken.

2) Marker placement (20 minutes)
After completing MVIC test, 22 Spherical-shaped retroreflective markers (10 mm diameter) with an adhesive surface will be placed to anatomical points at the shoulder and in your lower body to capture your movement.

3) Seven sequential walking tests (110 minutes)
You will wear seven different garments for each walking test.
The control garment will be a pair of snuggly fitting sport shorts (Figure 1a). Six treatments will include additionally wearing a 20lb (9.1 kg) Outer Tactical Vest (OTV) (treatment 1) with front and back ceramic plates (Figure1 1b).

4) Standing test on the pressure sensor mat (15 minutes)
After walking test is completed on the foot pressure sensor, you will stand still on the same pressure sensor mat for 30 seconds. You will repeat this standing test three times for each garment condition. Therefore, this standing test will take 2 minutes for each garment condition and the total time for the standing test will be about 15 minutes for seven garment conditions.

Between each garment treatment, you will be given 2~5 minutes of rest time to avoid possible fatigue. During the rest, you will sit in a chair with the carrying load removed. Then, the researcher will ask you about any perceived discomfort and fatigue that you might have experienced during the walking test. You will repeat walking tests with rest time for all seven garment treatments.

The study is designed to last approximately between 3 – 3.5 hours from taking the anthropometric measurement to completing all walking tests. All experiment procedures will be conducted in the Institute for Protective Apparel Research and Technology (IPART) lab, which is located in the Venture I building suite 109 located at the Oklahoma Technology and Research Park, 1110 S. Innovation Way, Stillwater, OK 74074.

RISKS OF PARTICIPATION:
There are no risks associated with this project, including stress, psychological, social, physical, or legal risk which are greater, considering probability and magnitude, than those ordinarily encountered walking in daily life. Although no subjects have encountered any problem in the previous experiments, it is possible that slight skin irritation from the marker and EMG placement or muscle soreness may occur. If, however, you begin to experience discomfort or stress during this project, you may end your participation at any time. In case of injury or illness resulting from this study, emergency medical treatment will be available at the OSU University health service (Monday-Friday from 8:00am - 5:00pm). No funds have been set aside by Oklahoma State University to compensate you in the event of illness or injury.

BENEFITS OF PARTICIPATION:
You may gain an appreciation and understanding of how research is conducted by using motion capture system.

CONFIDENTIALITY:
All information about you will be kept confidential and will not be used in any way such that data regarding an individual can be associated with that individual. Research records will be stored securely and only researchers and individuals responsible for research oversight will have access to the records. This information will be saved for three years for publication to occur. Results from this study may be presented at professional meetings or in publications. You will not be identified individually: Several photographs capturing your body below the neck will be taken for presentation or publication purposes with your permission. We will analyze data for the group as a whole. It is possible that the consent process and data collection will be observed by research oversight staff responsible for safeguarding the rights and wellbeing of people who participate in research.

COMPENSATION:
$70 cash will be provided to you after you complete all required tests.

CONTACTS:
You may contact any of the researchers at the following addresses and phone numbers, should you desire to discuss participation in the study and/or request information about the results of the study: Huiju Park, 431 HES, Dept. of Design, Housing and Merchandising, Oklahoma State University, Stillwater, OK 74078, (405) 612-2979; or Dr. Donna Branson, 405-269-3320. If you have questions about your rights as a

research volunteer, you may contact Dr. Shelia Kennison, IRB Chair, 219 Cordell North, Stillwater, OK 74078, 405-744-3377 or irb@okstate.edu

PARTICIPANT RIGHTS:

Your participation in this research is voluntary. There is no penalty for refusal to participate, and you are free to withdraw your participation in this project at any time.

CONSENT DOCUMENTATION:

I have been fully informed about the procedures listed here. I am aware of what I will be asked to do and the benefits of my participation. I also understand the following statements:

I affirm that I am 18 years of age or older.

I have read and fully understand this consent form. I sign it freely and voluntarily. A copy of this form will be given to me. I hereby give permission for my participation in the study.

________________________________ ____________________

Signature of Participant Date

I certify that I have personally explained this document before requesting that the participant sign it.

________________________________ ____________________

Signature of Researcher Date

I give permission to allow a photo of my body (below the neck) to be used for publication or presentation of the study's results.

________________________________ ____________________

Signature of Participant Date

Okla. State Univ.
IRB
Approved 11/1/11
Expires 10/14/11
IRB# HE1069-M

Appendix F

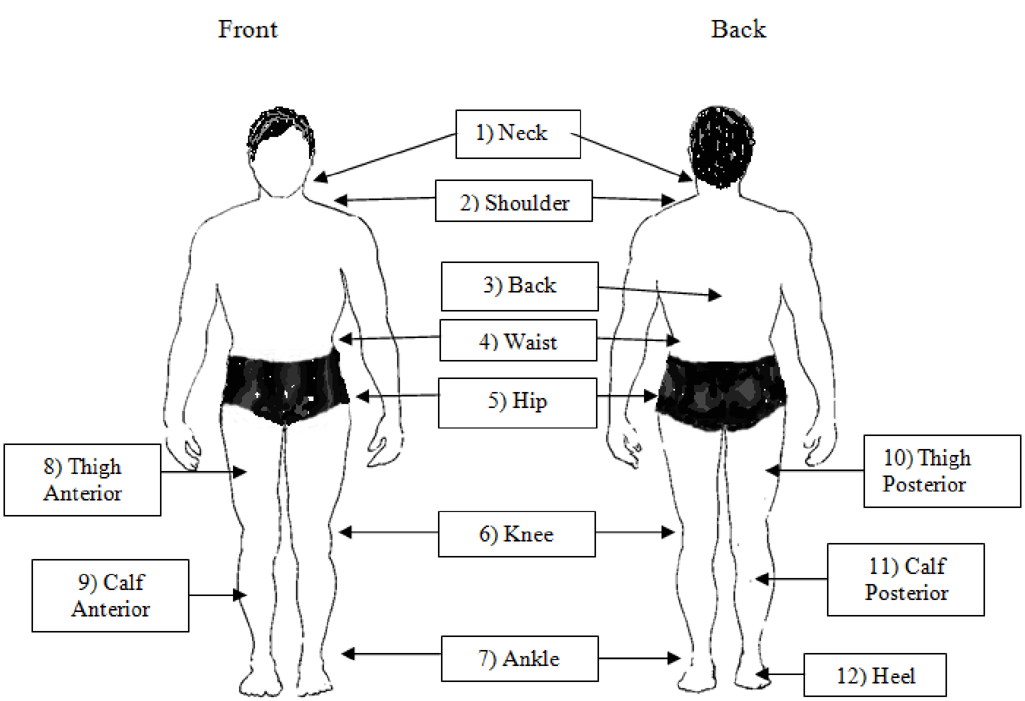
Wearer's Discomfort and Fatigue Ballot

Name: ____________________________ Treatment:____________________

I am going to ask you if you felt any discomfort and fatigue during the walking test that you just went through. I understand that you may feel a certain level of discomfort and fatigue from the experimental conditions such as markers and EMG probes attached to your body. However, I would like to ask you to focus on any discomfort and fatigue that you may have felt because of the garment and load carriage without any consideration of the experimental equipment.

Discomfort means any kind of unpleasant sensation and feeling that you experienced while you walked wearing the garments. The discomfort includes unpleasant pressure, rubbing/friction, resistance to movement, tightness and other possible psychological and sensorial unpleasant feelings.

1. Please look at the body area diagram on the desk. Were there any areas of the body in which you experienced discomfort during walking?



(1) Region (2) Type of discomfort and subject's comment

_________________________ ____

_________________________ ____

_________________________ ____

_________________________ ____

_________________________ ____

2. I am going to ask you about the fatigue you experienced during the walking test. Fatigue means a feeling and sensation of getting tired, weary, and even exhausted.

Were there any areas of the body in which you experienced fatigue during walking?

(1) Region (2) Subject's statement on fatigue

_________________________ ____

_________________________ ____

_________________________ ____

_________________________ ____

_________________________ ____

3. Overall, to what extent did this garment that you wore affect your walking?

No effect	Very slightly limited movement	Slightly limited movement	Limited movement	Severely limited movement
1	2	3	4	5

VITA

Huiju Park

Candidate for the Degree of

Doctor of Philosophy

Thesis: IMPACT OF BODY ARMOR AND LOAD CARRIAGE ON LOWER BODY MOVEMENT

Major Field: Apparel Design

Biographical:

Education:

Completed the requirements for the Doctor of Philosophy in the Department of Design, Housing, and Merchandising at Oklahoma State University, Stillwater, Oklahoma in July, 2011.
Completed the requirements for the Master of Science in the Department of Clothing and Textile Science at Yonsei University, Seoul, Korea in 2002.
Completed the requirements for the Bachelor of Science in the Department of Clothing and Textile Science at Yonsei University, Seoul, Korea in 2002.

Experience:

Research Associate at Oklahoma State University in Stillwater, Oklahoma, 2007-2011
Teaching Associate at Oklahoma State University in Stillwater, Oklahoma, 2010 Spring & 2011 Spring
Footwear and Sports Apparel Merchandiser at E.Land Co. (Puma Korea Division), Seoul, Korea, 2002-2007

Award:

Phoenix Award for Oklahoma State University Outstanding Doctoral Student at Oklahoma State University in Stillwater, Oklahoma, 2010

Professional Membership:

International Textile and Apparel Association

Name: Huiju Park   Date of Degree: July, 2011

Institution: Oklahoma State University   Location: Stillwater, Oklahoma

Title of Study: IMPACT OF BODY ARMOR AND LOAD CARRIAGE ON LOWER BODY MOVEMENT

Pages in Study: 125   Candidate for the Degree of Doctor of Philosophy

Major Field: Human Environmental Sciences – Apparel Design

Scope and Method of Study: The overall purpose of this study is to compare impacts that personal body armor and carrying loads have on lower body movement by using motion capture, Electromyography (EMG) and foot pressure technologies. Seven healthy male ROTC students participated in a human subject test. The independent variable was garment condition with seven levels. Seven treatment garments included 4 levels of weight (1/8 lb, 20 lb, 40 lb and 60 lb) with varying weight distribution. Treatment 1 was a 1/8 lb pair of snuggly fitting sports shorts. Treatment 2 (T2) was a 20 lb Outer Tactical Vest (OTV) in addition to wearing T1. Treatments 3, 4 and 5 (T3, 4 and 5) included wearing T2 and a 20 lb carrying load attached to three different OTV locations. Treatments 6 and 7 (T6 and T7) included wearing T2 and a 40 lb carrying load attached to the OTV in two different locations. Temporal and distance parameters of walking patterns, Range Of Motion (ROM)s, maximum joint angle (joint angle$_{max}$), peak EMG amplitude on four leg muscles, plantar pressure and contact area of the foot were measured as dependent variables while subjects walked barefoot wearing seven different garments. In addition, subjects' perceptions about ease of walking were also assessed by using a ballot with a 5-point Likert scale. Gait analysis and statistical analyses were used to identify changes in lower body movement while walking under the seven garment treatment conditions.

Findings and Conclusions: Significant garment effects were found for four temporal and distance parameters of walking (stance phase, swing phase, double support and stride length); six ROMs (pelvic obliquity, hip adduction-abduction, pelvic tilt, hip flexion-extension, pelvic rotation and hip rotation); two joint angle$_{max}$ (pelvic intrarotation and pelvic extrarotation); peak EMG amplitude on the rectus femoris; and peak plantar pressure at the forefoot and rearfoot, average plantar pressure and change in contact area. Subjects reported perceiving increased limitation in lower body movement, discomfort and fatigue, as weight of garment and carrying load increased.

ADVISER'S APPROVAL: Dr. Donna Branson