EFFECTS OF LOAD CARRIAGE

AND HIGH-HEELED SHOES

ON SPINAL MOTOR CONTROL

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Certificate of Originality

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WANG Chao

Abstract

Load carriage has been identified as a potential risk factor for low back pain. Previous studies of the effects of load carriage on spine were conducted under quasi-static conditions. In upright stance, it was shown that repositioning consistency and muscle activity of the lumbar spine were significantly reduced during load carriage. Thus, the effect of load carriage on spinal motor control and its possible association with the cause of low back pain have been our concern. On the other hand, high-heeled shoes are commonly used nowadays as a f ashion for ladies. It was shown that back muscle activity was increased in wearing high-heeled shoes. It was thought that whether high-heeled shoes could be used to counteract the effect load carriage on t he spine or not. The objectives of the current study are to investigate the effects of load carriage on spinal motor control under dynamic condition and to explore the possibility of using high-heeled shoes to counteract the effect of load carriage. The study was divided into two phases. In phase I, dynamic system theory was applied to study the movement coordination of the lumbar spine relative to pelvis under different weights of load carriage (0, 5, 10 and 15% of body weight (BW)). In phase II, the combined effects of load carriage (0, 5, 10 and 15% BW) and high-heeled shoes (0, 2 and 5cm heel height) were investigated.

Eight male and eight female healthy volunteers participated in phase I and another twelve female healthy subjects participated in phase II of the study. In both phases I and II, functional reaching distances (FRD) of each subject under different conditions (i.e. with and without load carriage at different heel heights) were determined by a standard

functional reaching test. Afterwards, the subject was asked to perform three consecutive and continuous movements which consisted of symmetric forward reaching to a midline target located at shoulder height and returning to the upright standing posture with feet at shoulder width, shoulders in 90° flexion and fully extended elbow. The target distance for each subject was standardized to 50% of individual's FRD.

Reflective markers were affixed to the participants' spine, pelvis and thigh and their coordinates were captured by a motion analysis system (Vicon Nexus, Oxford Metrics, Oxford, UK) during the entire motion. Kinematics of the lumbar spine, pelvis and the thigh in sagittal plane were determined. The initial upright posture, the repositioning consistency of the upright posture and lumbar movement ratio were also determined for each condition. Based on the dynamical systems theory, two parameters, namely, mean absolute relative phase (MARP) and deviation phase (DP) were calculated for studying the movement coordination between lumbar spine and pelvis. The results were analyzed using repeated measure analysis of variance (RANOVA) with level of significance set at $p=0.05$.

It was found that the initial upright posture was not significantly affected by load carriage. In comparison with the unloaded condition, repositioning consistency of lumbar spine was found to be significantly decreased during carrying load even the weight was only 5%BW. FRD was found to be significantly decreased with increased load carriage and heel height. Load carriage was also found to induce significant increase in lumbar movement ratio, MARP and DP. However, the effects of heel-height on these three parameters were opposite, high-heeled shoes were found to decrease lumbar movement ratio, MARP and DP. Besides spinal motor control was significantly affected under quasi-static situation, dynamic spinal motor control was also significantly affected by load carriage. Though there was no interaction between loading and high-heeled shoes, the combined effects may counteract the adverse effects of load carriage to some extent. Also other pragmatic approaches should be considered to elucidate the adverse effects of load carriage on the spine.

Publications from the Thesis

Conference Presentation

C. WANG, D.H.K. CHOW and A. LAI. Effects of load carriage on spine dynamical properties under sudden perturbation. Proceedings, Biomedical Engineering International Conference, Hong Kong, A-5, 2-5 Nov, 2010.

C. WANG, D.H.K. CHOW and A. LAI. Effects of load carriage on trunk motor control. Proceedings, 1st International Congress Scientific Testing of Orthotic Devices, Aix les Bains, France, p25, 23-25 Mar, 2011.

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Chapter 1 Introduction

It has been demonstrated that loading on s pine could adversely affect a number of physiological parameters, such as muscle activity (Hong et al., 2008; Motmans et al., 2006) and cardiopulmonary function (Li et al., 2003). Moreover, spine curvature (Chow et al., 2007), spine repositioning consistency (Chow et al., 2007; Chow et al., 2010) and postural stability (Goh et al., 1998; Palumbo et al., 2001; Pascoe et al., 1997) were also affected during load carriage. Load carriage has been proposed to be associated with low back pain. However, there was no obj ective experimental data that could support this causal relationship. In children, the deficit of spine motor control was found with reduced spine repositioning consistency (Chow et al., 2007). In adults, similar biomechanical and physiological changes were observed during load carriage. However, it is still uncertain whether this was a natural body adaptation to the load carried or beared any relationship to increased risk of back injury or back pain. Moreover, it was demonstrated that erector spinae relaxed during posterior load carriage (Motmans et al., 2006). Although the loading acting on spine may be reduced during posterior load carriage, it is still questionable whether the observed reduced spinal motor control under quasi-static condition will increase the risk of spinal injury.

A deeper understanding of the effects of load carriage on spinal motor control in particular under dynamical situations will be useful for filling this knowledge gap. Therefore, one of the objectives of this study was to apply dynamical systems theory to investigate the effects of load carriage on spinal motor control under dynamical situations. In addition, the activity of erector spinae was decreased during load carriage, while high-heeled shoes were found to be able to activate erector spinae. Thus high-heeled shoes were proposed to be a possible means to counterbalance or minimize the effects of load carriage here. Hence the second objective of this study was to test whether there was any interaction effect on spinal motor control between high-heeled shoes and load carriage.

Chapter 2 Literature Review

2.1 Low Back Pain

Low back pain (LBP) is a common musculoskeletal problem and over 80% of the world's population suffered from LBP at some point in their lives (Waddell, 1996). The exact cause of low back pain is still unknown, but a previous history of LBP has been found to be associated with future onset of LBP (Jones & Macfarlane, 2005). Therefore, prevention of LBP is an important issue and numerous studies have been conducted to identify the risk factors for LBP.

Many daily load carriage users reported that they have low back discomforts as well as LBP. T hese include schoolchildren, adolescents, soldiers, college students and industrial workers (Birrell & Haslam, 2009; Sheir-Neiss et al., 2003; van Vuuren er al., 2007). People subjectively think that load carriage is a risk factor for LBP. Although it was found that backpack carried by adolescents with back pain was significantly heavier than that of those without back pain (Sheir-Neiss et al., 2003), there was little experimental data that could confirm this potential casual relationship. Recently, it was demonstrated that spinal proprioception (Chow et al., 2007) and balance control were significantly affected by load carriage and these effects were found to be associated with risk of spine injury and falling. It was also reported that individuals with LBP exhibited longer trunk muscle response latencies than healthy controls when confronted with sudden loading (Radebold et al., 2000). An impairment of motor control has been proposed to be a possible cause of low back injury and recurrence of LBP (Cholewicki et al., 2005). In order to have a deeper understanding of the effects of load carriage on the spine, a comprehensive literature review is conducted.

2.2 **Load Carriage**

2.2.1 **Daily Use of Load Carriage**

Load carriage is common among schoolchildren, adolescents and adults for daily transferring personal belongings, books and stationeries, laptops to and from workplaces or schools. Many people are also required to take load carriage in their daily life, such as postmen, soldiers, students and recreational hikers.

In posterior load carriage, additional load was placed on s pine directly through the shoulder straps (Negrini & Carabalona, 2002). There were suggestions that weight and duration of load carriage might be associated with LBP. Korovessis et al. (2005) also suggested that the weight of load carriage may be associated with musculoskeletal deformities such as scoliosis, kyphosis and lordosis. Although there was evidence for the association between load carriage and musculoskeletal disorders, it is still unclear whether there is any casual relationship.

2.2.2 **Effects of Load Carriage**

The effects of load carriage on body posture, physiological performance, gait pattern and muscle activities have been widely investigated.

Body Posture

An increase in trunk forward lean was found to be associated with the weight of posterior load carriage (Chow et al., 2007; Hong & Cheung, 2003). The amount of head extension was found to be increased with backpack weight (Chow et al., 2006; Vacheron er al., 1999). It was thought that the extension was a compensation for the increased trunk forward lean to maintain eye gazing. The increased forward head posture combined with increased head extension during backpack loading might cause an increase in shearing stress in the cervical spine and so the stain at the cervical intervertebral discs. Prolonged adoption of this protracted head posture might increase the risk of neck pain (Grimmer et al., 1999). The effect of backpack carriage on thoracic kyphosis was examined in several studies. Vacheron et al. (1999) and Chow et al. (2007) found that thoracic kyphosis was flattened during backpack carriage. Chow et al. (2007) found that the reduction in thoracic kyphosis was only significant in the upper thoracic region rather than the lower region. The decrease in thoracic kyphosis might due to contraction of the trapezius muscles (Hong et al., 2008). Lumbar lordosis was found to be reduced during backpack carriage (Chow et al., 2007; Vacheron et al., 1999). The lumbar lordosis tended to decrease with increasing backpack load (Chow et al., 2007). The decreased muscle activity of erector spinae was thought to have an important role in maintaining trunk posture (Motmans et al., 2006) and it might be the cause of the lumbar lordosis reduction. It was also believed that the decrease in lumbar lordosis was a consequence of a retroversion movement of the pelvis which led to horizantalization of the superior S1 level (Vacheron et al., 1999).

In summary, body alignment and spine curvature were found to be deviated from normal upright posture during backpack carriage in many studies. As normal upright posture allows the body to maintain balance with minimal muscular effort, the postural deviation during backpack carriage might increase the internal energy expenditure (Kendall, 2005) and might also increase the stress and strain on the spine (Kendall, 2005).

Physiological Effects

Apart from the body posture and spine curvature changes, it was also demonstrated that physiological performance was affected by posterior load carriage. Increases in heart rate, blood pressure and energy expenditure were found during walking with load carriage (Hong & Brueggemann, 2000). The pulmonary function was also noted to be affected when carrying posterior load carriage (Chow et al., 2005). A significant decreased forced expiratory volume (FEV1) and forced vital capacity (FVC) was shown when a heavy load carriage weight with 20% BW to 30%BW was carried (Lai & Jones, 2001). The changes of FEV1 and FVC were further shown to be associated with load weights. These two parameters were found to be decreased significantly with increase of load weight (Chow et al., 2005).

Several studies have investigated the effects of load carriage on gait performance. Significant differences in walking speed, cadence, stride length, stride frequency, swing duration and double support time were observed with increasing load (Chow et al., 2005; Hong & Brueggemann, 2000; Pascoe et al., 1997). However, different observations were reported by some studies. Goh et al., (1998) observed that walking speed and stride length remained unchanged in normal male adults when carrying backpack with 0%BW, 15%BW and 30%BW. Hong and Cheung (2003) also found no significant differences in stride length, cadence, velocity, single support time or double support time when carrying posterior load carriage of 0% BW, 10% BW, 15% BW and 20% BW. The differences in findings among these studies may be due to gender and age of the participants or the sample size.

The muscle activity patterns during load carriage have also been widely studied. The carrying load was balanced either by the relaxation of the back muscles or the contraction of the abdominal muscles (Motmans et al., 2006). A significant increase in activation of rectus abdominis and obliques externus abdominis was found with increasing loading weight, while the muscle activation of trapezius pars descendents, rectus femoris and biceps femoris were affected minimally by load carriage (Devroey et al., 2007).

2.2.3 **Load Carriage and Low Back Pain**

Various studies have been conducted on the effects of extra loading induced to the spine by load carriage and the relationship to low back pain. Goh et al. (1998) found an increase in the peak lumbosacral force by 27% and 30% while walking with load of 15%BW and 30%BW compared with the no l oad condition. The spine anatomical structure was also reported to be affected by external load (Kimura et al., 2001). It was found that load carriage significantly narrowed the lumbar dural sac and changed the intervertebral angle. The changes of interveterbral heights and angles in adult population were measured in supine position using Magnetic Resonance Imaging (MRI). However, the relation between load carriage and low back pain was not direct. This conclusion was also in agreement with the study by (Viry et al., 1999). It was suggested that fatigue and time spent on load carriage were associated with back pain. These studies indicated that there was potential relationship between load carriage and back problem. However, the exact relationship is still not fully understood.

2.3 High-heeled Shoes

Since loading on spine could affect a number of physiological parameters adversely, many studies have been conducted in an attempt to minimize or counterbalance the effects of loading. Different carrying methods have been considered, such as anterior, posterior as well as symmetrical and asymmetrical load carriage. In this study, the use of daily used high-heeled shoes was investigated for possible counterbalancing the effects of load carriage. The rationale and the effects of high heel shoes are elaborated below.

2.3.1 Daily Use of High-heeled Shoes

Since the 17th century, women have worn high heel shoes. In this modern society, many women wear high-heeled shoes in both professional and social settings. Recent evidence showed that 59% of women wear high heel shoes for 1 to 8 hours per day. It has previously been suggested that wearing high heel shoes may have adverse effects on the musculoskeletal system.

A number of studies have investigated the gait pattern with high-heeled shoes using kinematic, kinetic and physiological techniques. These studies have indicated that high-heeled gait is less energy efficient than low-heeled gait and can increase rate of fatigue, decrease reflex and voluntary movement response rate as well as alter muscle onset time and muscle strength. However, most of the previous studies focused on the effects of high-heeled shoes on lower limb rather than the spine.

2.3.2 **Effects of High-heeled Shoes on Spine**

Trunk Muscle Activity

Effects of high-heeled shoes on muscle activity during gait have been widely studied. Most of these studies have concentrated on the electromyographic (EMG) activity of lower limb muscles. Only a few studies examined the effects of heel lifts on the back muscles. These limited numbers of studies suggest that ambulating with an increased heel height alters the onset timing of the erector spinae muscles. For example, Bird et al. (2003) observed significantly earlier in erector spinae activity during gait with bilateral heel lifts indicating the foot wedging can produce measurable changes in the timing of muscle activity within the back and pelvis muscles during gait. These effects and the consequences of changes in muscle activity patterns of the lumbar musculature may be clinically significant because it has previously been reported that a small but prolonged increase in EMG activity of the back muscles may lead to chronic overload and fatigue of the muscles. Recently, Mika et al. (2012) found that the erector spinae muscles exhibited an increase in EMG activity in association with an increase in heel height. From a clinical perspective, increased lumbar erector spinae muscle activity could exacerbate muscle overuse and lead to low back problems. Mika et al. (2012) conducted another study to evaluate the changes of EMG in cervical paraspinal muscle during gait in high heel shoes. Higher EMG activity cervical paraspinal muscle was noted in high heel shoes in comparison to walking without shoes. The prolonged wearing of high-heeled shoes by individuals without neck pain is not safe for their spine and may lead to chronic paraspinal muscle fatigue (Mika et al., 2011).

Body Posture

The effects of positive heel inclination on postural alignment of the head, spine, pelvis, and knees have been studied (Franklin et al., 1995). It was found that positive heel inclination of subjects caused significantly lower anterior pelvic tilt, lumbar lordosis, and sacral base angles when compared with zero heel inclination. Clinically, patients with low back pain may be affected by high heel usage because of the reduction of the normal lumbar lordosis. Comparison of barefoot and high-heeled stance showed that the wearing of high heels caused lumbar flattening, a backward tilting pelvis and a posterior displacement of the head and thoracic spine (Opila et al., 1988).

2.4 Motor Control

Motor control of spine was thought to be a critical factor related to low back pain. The deficit of spinal motor control during load carriage or high-heeled shoes was the potential explanation to the cause of low back pain.

Movement is a critical and essential to our daily activity and ability to survive. The field of motor control is to study the nature of movement and its control. Motor control is usually defined as the ability to regulate or direct the mechanisms essential to movement.

Movement can be considered to be emerged by the interaction among three factors: the individual, the environment and the task (Shumway-Cook & Woollacott, 2007). Movement is organized under the demands of the task and the environment. The individual then generates movement to meet the demands of the task under a specific environment. The performer’s capacity to meet the task and environmental demands determines the person’s functional capacity.

Motor control is determined by the input from the somatosensory, visual, and the vestibular systems (Magnusson et al., 2008). These systems provide the stimuli to initiate movement and feedback to modulate the movement corporately. The somatosensory system provides spatiotemporal information of the body and limbs. This includes information of muscle tension and length, joint angles as well as joint velocities. Although proprioception is provided by the somatosensory system, the visual system also plays an important proprioceptive role. Finally, the vestibular system provides information regarding the head position and the changes in the direction of head movement. The processes involved in motor control are complex and can be divided as perception, motor planning, motor execution and feedback phases.

2.4.1 **Process of Motor Control**

Perception

The process of perception starts with an object in the real world which is termed as the distal stimulus. Through light, sound or other physical process, the objects stimulated the body's sensory systems. The input energy was transformed into neural activity by the sensory organs through a process called "transduction". This kind of neural activity is named the proximal stimulus (Shumway-Cook & Woollacott, 2007). These neural signals are then transmitted to the brain and processed. The resulting mental recreation of the distal stimulus is a perception. A simple example of perception would be an individual who gazes at a ball. The ball itself is a distal stimulus. When the light of the ball enters one's eyes and stimulates the retina, that stimulation is the proximal stimulus. The image of the ball reconstructed by the brain is the percept.

Sensory Integration

The next process is to integrate the sensory input received by different organs. These include the senses of vision, audition, tactile stimulation, smell and so on. It is important that the information of different sensory modalities is relatable. Sensory integration is usually defined as the neurological process that organizes sensation from the environment and one's own body for controlling the body effectively under the constraints due to the environment and the task. Through sensory integration, the brain can relate all sensory inputs into a coherent percept. Sensory integration is necessary for almost every activity because the combination of multiple sensory inputs is essential for us to comprehend our surroundings.

Motor Execution

After the sensory integration process, the motor unit will execute the expected movement. Muscles will generate sufficient tension for the purpose of the planned posture and movement. Sometimes this was completed by the cooperation of several muscles rather than a single independent muscle. This is a result from both the musculoskeletal properties and neural activation of the muscles.

Feedback

After the execution of the motor programming, some feedback signals are necessary for correcting and revising the error between the actual movement trajectory and the planned ideal trajectory. The feedback includes all the sensory information that is available. This is also called a response-produced feedback and is usually further divided into two subclasses, namely, intrinsic feedback and extrinsic feedback (Schmidt & Lee, 2005). Intrinsic feedback comes to the individual through various sensory systems. This includes information such as visual information concerning whether a movement was accurate or somatosensory information concerning the position of the limbs as one was moving (Schmidt & Lee, 2005). Extrinsic feedback is the information for supplementing the intrinsic feedback. For example, verbal feedback is given to a patient in clearing an object while walking. Extrinsic feedback can be given concurrently with the task and in addition, at the end of the task.

Central Nervous System

The central nervous system (CNS) is the main part of the nervous system for integrating the received information and coordinating the activity of all parts of the bodies. It consists of the majority of the nervous system and includes the brain and the spinal cord. Together with the peripheral nervous system, it has a fundamental role in the control of behavior.

2.4.2 **Quantification of Spinal Motor Control**

Spine is an important structure of human body to bear loads, allow movement and so on. Several different methods have been applied to quantify spinal motor control.

Proprioception

Proprioception, also named repositioning consistency, is the sense of the relative position of neighboring parts of the body. It is the sense that indicates whether the body is moving with the required effort, as well as the various parts of the body are located in relation to each other. Trunk proprioception has been used as an evaluation parameter of spinal motor control in many studies. Chow et al. (2007) examined spine proprioception of schoolboys. Subjects were asked to keep in a relaxed upright stance with the arms at the sides and the feet spaced apart at a comfortable distance. The study participants were instructed to keep their gaze on a target at eye level 2m directly in front of them and the spine curvature were recorded for 3s using a motion analysis system. The participants were then instructed to walk around a 6m loop, stand back to the feet positions marked on the floor and gaze at the target in front of them again and the spine curvature were measured again for 3s. This process was repeated six times and the standard deviation of the six measurements was determined as the repositioning consistency.

Recently, Lee et al., (2010) developed a method for quantifying proprioception in which motion perception threshold, passive repositioning and active repositioning were measured. For measuring motion perception threshold, a stepper motor was used to rotate the lower body at 0.1°/s away from the neutral position. The subjects were asked to press a handheld button once they perceived a change in position and to report the direction of motion. The trials were recorded if the subject reported the direction correctly. For measuring passive repositioning, the stepper motor moved the subjects' lower bodies 15° a way from the neutral at 2.2°/s. Once 15° w as reached, the motor briefly paused and started to return toward the neutral position at 1.0°/s. The subjects were asked to press a button when they perceived they were back to the neutral position. The process for measuring active repositioning was similar to that of passive repositioning except that once the motor reached 15° the clutch was disengaged and the subjects had to actively reposition their lower bodies to the perceived neutral position. When the subjects perceived that they had returned to the neutral position, they had to press a button and the angle was recorded. The subjects were given 2 trials to get familiar with the setup in each plane of motion prior to data collection. For lateral and axial planes of motion, 4 trials were conducted in each direction and the data from left and right directions were combined. In the study by Lee et al. (2010), 5 trials of measurements were taken for flexion and extension directions as they could not be combined.

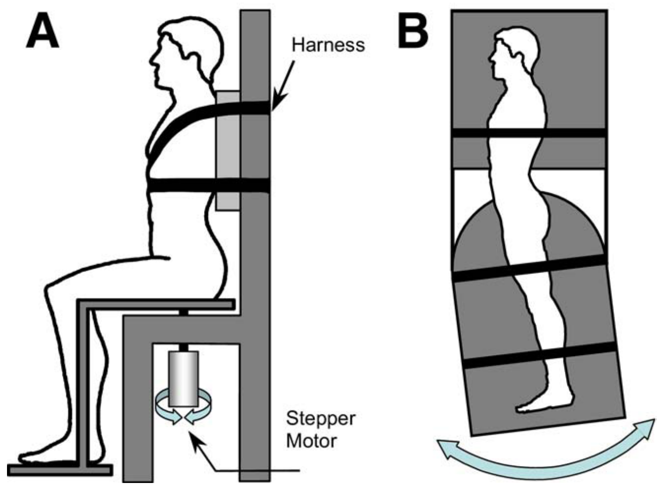


Figure 2.1 The apparatus for assessing proprioception in (A) axial rotation and (B) flexion and extension. For lateral bending, the same setup as flexion and extension was used, but the subjects were lying in a supine position (Lee et al., 2010)**.**

Although proprioception is mainly provided by the somatosensory system, the visual system plays a proprioceptive role. Proprioception was found to be greatly decreased without visual input (Silfies et al., 2003).

Muscle Activity

Muscle is the execution part of the motor control system. Muscle strength is defined as the ability to generate sufficient tension in a muscle for the purpose of posture and movement (Schmidt & Lee, 2005; Smidt & Rogers, 1982). EMG signals are typically measured by surface electrodes. Main trunk muscles include internal oblique, external oblique, rectus abdominus, lumbar erector spinae and lumbar multifidus. Muscle activity, muscle onset and offset time were recorded to represent the muscle execution function in previous studies (Al-Khabbaz et al., 2008; Cholewicki et al., 2005; Hong et al., 2008; Motmans et al., 2006). Moreover, it was found that muscle recruitment pattern was changed and trunk muscle reflex was delayed in patients with low back pain (Cholewicki et al., 2005; Reeves et al., 2006).

Spine Stability

Stability is one of the most fundamental concepts to characterize and evaluate any system. Spinal stability refers to the ability of the spine to bear loads, allow movement, and at the same time avoid injury and pain. In assessing spinal stability, some consider only the mechanical contributions of passive anatomical muscular system while clinicians assess stability from the symptomatic standpoint (Adams, 2007; Reeves, et al., 2007). Postural stability has been used as an overall human motor control assessment and commonly assessed by measuring the trajectory of center of pressure (COP).

A spine stability test was proposed by (Reeves et al., 2006), which was called an unstable seated balance test. Subjects were placed on a seated equipped with leg and foot supports to prevent any lower body movement (Cholewicki et al., 2000). In this test, the subjects were instructed to sit on a surface with a 30cm diameter polyester support which was placed on a force platform at the edge of a table (Figure 2.2). The subjects were asked maintain their balance while seated with arms crossed. In case of loss balance, a safety railing around the force plate was attached for protection.

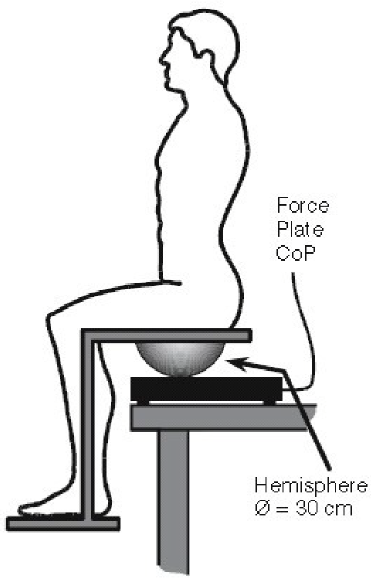


Figure 2.2 Subject positioned in the unstable sitting apparatus. Centre of pressure (COP) movement was recorded by the force plate located beneath the hemisphere (Reeves et al., 2006)**.**

Another method for assessing trunk stability through spine mechanical properties was proposed by(Cholewicki, Simons, & Radebold, 2000). Human trunk was modeled as a second-order mass-spring-damper system, which involved a mass m (in kg), a spring constant k (in N/m) and a damping coefficient B (in N•s/m).

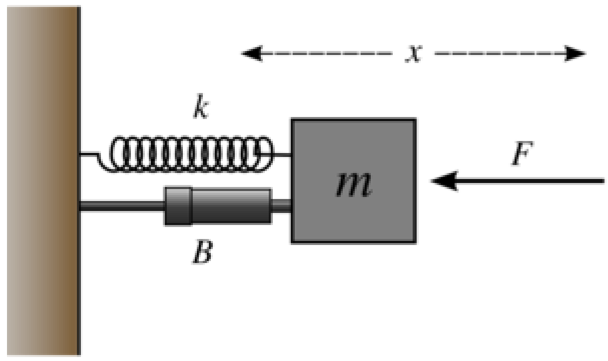


Figure 2.3 Human trunk was modeled as a second-order mass-spring-damper system (Cholewicki et al., 2000)**.**

It was shown that people with LBP have increased estimated effective trunk stiffness but with decreased damping (Hodges, van den Hoorn, Dawson, & Cholewicki, 2009) which was thought to be a sign of deficit of trunk motor control. As a stiffer system will be displaced less than a compliant system, the causes of increased trunk stiffness were proposed to be due to an increase of trunk muscle co-activation with intention to improve spinal stability with reduced intersegmental displacement so as to minimize possible risk of injury. However, as spinal loading would also be increased with increased trunk muscle activity and this might adversely contribute to ongoing LBP (Hodges et al., 2009). A well-damped system will return to equilibrium position with very few oscillations. A poorly damped system would take longer and undergo more oscillations to reach the equilibrium. With regard to the dynamic stability of the spine, damping could dissipate kinetic energy and it is an important feedback component for preventing possible spinal injury during dynamic motion.

Movement Coordination

Silfies (2010) found that trunk movement variability of patients with LBP was increased. In a dynamic environment, such variability may permit people to move in reaction to the demands of task and environment, in a more efficient and stable way over time. Therefore, movement variability could be viewed as healthy and essential for optimal flexibility and stability. However, a significant reduction or increase in movement variability could also represent abnormal states. Increased variability could result from a performer's inability to discover a more stable motor solution following environmental perturbations or altered task demands. Additionally, greatly decreased variability might be a sign of pathological state with limited movement options.

In a coordinate movement, multiple joints and muscles are activated at the appropriate time and with the correct amount of force so that smooth, efficient, and accurate movement occurs. Thus, the essence of coordination is the sequencing, timing and grading of the activation of multiple muscle groups. Because of the synergistic nature of coordination, the capacity to generate force in an isolated muscle does not predict the ability of that muscle to work in concert with others in a task-specific way (Giuliani, 1991). For investigating the movement coordination of the trunk, different parts of the whole spine such as thoracic and lumbar or the lumbar and pelvis should be considered.

The relationship between the movements of the lumbar spine and hip has been widely studied. The commonly used tasks included forward-backward bending, lateral bending and sit-to-stand task. Angle-angle plot and cross-correlation of angular displacements of body segments were used to examine the movement coordination of the spine relative to the hip (Lariviere et al., 2000; Lee & Wong, 2002). Silfies et al. (2009) applied a dynamical system theory (DST) approach to characterize movement of lumbo-pelvic region between healthy people and patients with LBP during a reaching task. DST has been used to study movement coordination and stability of coordination within the human body (Kurz & Stergiou, 2004). It was found that the movement coordination between lumbar spine and pelvis was more out of phase in patients with LBP (Silfies et al., 2009).

A wide variety of measures such as joint kinematics, joint moments and electromyography have been used in the literature to define the organization of the neuromuscular system. These approaches have provided useful scientific information that has advanced our understanding of the organization of the system for healthy and pathological movement patterns. However, it becomes an overwhelming task to determine which biomechanical variables actually capture the state of the neuromuscular system. The use of the Dynamical systems theory (DST) allowed the behavior of the neuromuscular system be expressed theoretically in a low-dimensional term (i.e. one variable) so as to offer a better way to gain scientific information on the organization of the system in performing functional movement patterns.

2.5 Dynamical Systems Theory & Phase Portrait

2.5.1 Dynamical Systems Theory

Effective organization of the multiple degrees of freedom present in the neuromuscular system has been theoretically proposed as a necessity for healthy functional movement patterns (Turvey, 1990). The inability of the neuromuscular system to synergistically orchestrate the many degrees of freedom would result in pathological movement patterns. Traditionally, different biomechanical tools have been utilized to define the dynamic organization of the neuromuscular system. Many researchers used a wide variety of measures such as joint kinematics, joint moments, and electromyography to define the organization of the neuromuscular system (Birrell & Haslam, 2009; Chow et al., 2010; Kim et al., 2011; Li et al., 2003; Snow & Williams, 1994). These approaches have provided useful scientific information that has advanced our understanding of the organization of the system for healthy and pathological movement patterns. However, it becomes an overwhelming task to determine which biomechanical variables actually capture the state of the neuromuscular system. The application of the Dynamical Systems Theory (DST) was to express the behavior of the neuromuscular system in a low-dimensional term (i.e. one variable) and to select the proper biomechanical variables that capture the organization of the neuromuscular system.

According to the principles of DST, movement patterns arise from the synergistic organization of the neuromuscular system based on m orphological factors (i.e. biological constructs), biomechanical variables (i.e. Newton Laws), environmental factors (i.e. spatial and temporal configuration of events), and task constraints (e.g. walking at slow or fast speeds) (Lockman & Thelen, 1993; Thelen & Ulrich, 1991). Therefore, the generation of movement pattern is multifactorial and that movement involves the coupling of the multiple degrees of freedom present in the human body. Movement patterns are then the results of the individual muscles and neuropathways collectively working together to achieve a functional outcome that meets the constraints of the system. Such coordinative structures in the extremities often span more than one joint (Kelso, 1995).

Slight variations in the way the degrees of freedom are coupled together in the coordinative structure provide a rational as to why no two steps are exactly alike during gait and why inter-subject exist for completing the same movement pattern (Clark & Phillips, 1993). Dynamical systems theory suggests that variations in the movement patterns are attributable to the neuromuscular system's response to global (changes in environment or task) and local perturbations (e.g., joint flexibility and proprioception) (Lockman & Thelen, 1993; Thelen & Ulrich, 1991). This would suggest that variations in the way the neuromuscular system is organized may be related to health. In other words, abnormal movement patterns may be due to an inability of the coordinative structures to organize the degrees of freedom in an effective way to adapt to perturbations experienced.

2.5.2 **Phase Portrait**

One can view the behavior of a dynamic system as a differential equation in which the changing state of the system is a function of a state vector (Arbarbanel, 1996). Although the differential equation for the system is typically unknown, plotting the current state of the system versus its rate of change could be used to understand the behavior of the dynamic system (Clark & Phillips, 1993). This type of plot is referred to as a phase portrait. A phase portrait provides a qualitative picture of the organization of the neuromuscular system. Changes in the configuration of the phase portrait provide initial insight into the control mechanisms (Winstein & Garfinkel, 1989).

The phase angle of the phase portrait trajectory quantifies the behavior of the involved segment and is used to calculate relative phase. To calculate the phase, the phase portrait trajectories are transformed from Cartesian (x,y) to polar coordinates, with a radius r and phase angle θ (Clark & Phillips, 1993; Scholz & Kelso, 1989). The angle formed by the radius and the horizontal axis is the phase angle of the trajectory (Figure 2.4).

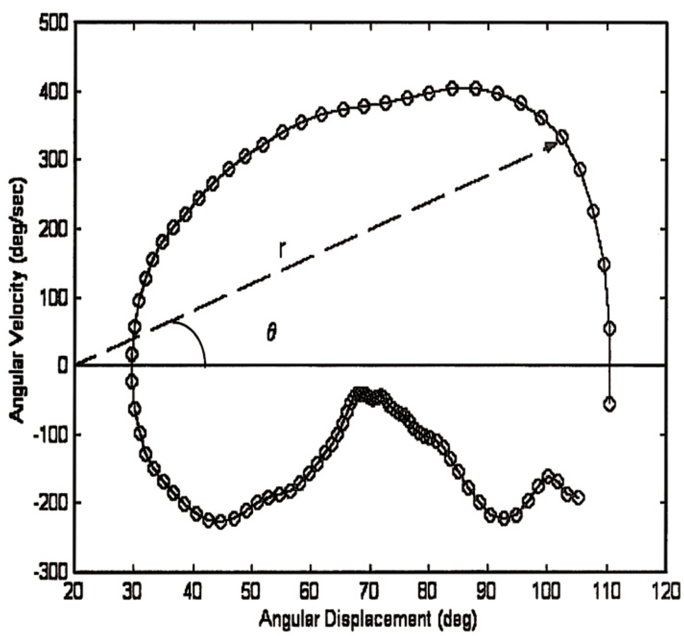


Figure 2.4 Shank phase angle (Kurz & Stergiou, 2004)**.**

Figure 2.4 displays a phase portrait for a segment during gait. The angle formed between the x-axis and the vector r is called the phase angle. This angle quantifies where the trajectory is located in the phase portrait as time progresses. As indicated in this figure, positive phase angle are calculated if the trajectory is within quadrant 1, and negative phase angles are calculated if the trajectory is within quadrant 4.

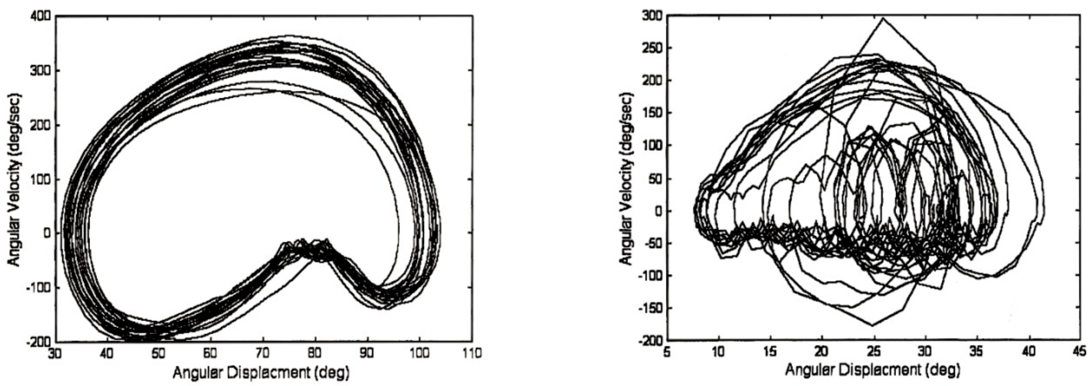


Figure 2.5 Shank-thigh phase portrait during gait (Left) Normal healthy gait (Right) Parkinsonian gait (Kurz & Stergiou, 2004)**.**

When multiple gait cycles are plotted on the same phase portrait (Figure 2.5), the amount of variability in the path of the trajectory can be used to qualitatively assess the stability of neuromuscular system (Clark & Phillips, 1993). Slight variations in the trajectories are due to the neuromuscular system's response to global and local perturbations experienced during the gait cycle (Clark & Phillips, 1993). Such flexibility allows the neuromuscular system to maintain a stable and proficient movement pattern.

However, excessive variability has been associated with instabilities in the behavior of the neuromuscular system (Clark & Phillips, 1993). Such instabilities are evident in the Parkinsonian gait portrayed in figure 2.5(Right).

DST emphasizes the identification of a low-dimensional parameter that defines the dynamic state of the neuromuscular system (Barela et al., 2000). This variable is referred to as an order parameter. The order parameter compresses the multiple degrees of freedom contained in the movement pattern into one value. Previous work has demonstrated that the relative phase relationship between the lower extremity segments (i.e. shank-thigh) is an order parameter that defines the collective state of the neuromuscular system during gait (Barela et al., 2000; Clark & Phillips, 1993; Diedrich & Warren, 1995; Stergiou et al., 2001).Selection of relative phase as an order parameter is based on the facts that the segments of the lower extremity conform to a limit cycle attractor, that relative phase variability increases prior to behavior transitions, and the relative phase variability decreases once a new behavior is selected. Therefore, on the basis principles of DST, relative phase captures the dynamic organization of the neuromuscular system in a low-dimensional term. Since relative phase encompasses angular displacement and velocity within one variable, some have argued that relative phase provides a better measure of the organization of the neuromuscular system than other biomechanical measures (Barela et al., 2000; Kelso, 1995). This rationale is supported by logical evidence of receptors in the joint that are responsive to changes in both displacement and velocity (McCloskey, 1978). Since relative phase accounts for such biological properties as one variable, it has a distinct advantage for determining the organization of the neuromuscular system. Additionally, Barela (2000) reported that relative phase provided a b etter measure of changes in the organization of the neuromuscular system than traditional biomechanical measures (i.e. joint angular displacement).

2.6 **Research Question**

In summary, backpack carriage could induce extra loading on s pine and change the trunk posture, as well as body performance. These changes may be further associated with LBP. A deficit in spinal proprioception has been associated with spinal disorders, and poorer repositioning ability has been reported during load carriage. However, these effects were investigated under quasi-static conditions, it is imperative to extend our standing to dynamical movement situations. Dynamical system theory has been applied to study spinal movement coordination and stability during load carriage. It is hypothesized that load carriage would also deficit the dynamical movement. Moreover, many studies have been conducted to try to minimize or counterbalance the effects of loading. Different carrying methods have been considered, such as anterior or posterior as well as symmetrical or asymmetrical carriage. Since high-heeled shoes was found to be able to active erector spinae whose activity was decreased during load carriage, in this study, daily used high-heeled shoes were employed as a t ool to try to counterbalance the effects of load carriage. The objectives of the current study are therefore to investigate the dynamical effects of load carriage on spine, and explore the possibility of applying high-heeled shoes to counterbalance the effects of load carriage.

Chapter 3 Methodology

3.1 Experimental Design

The study was divided into two phases. In the first phase of the study, the effects of load carriage on spinal motor control were evaluated under different carrying weights. In the second phase of the study, the possibility of using high-heeled shoes as a strategy for counteracting the effects of load carriage was explored.

In the first phase of the study, the subjects' spinal motor control was assessed by a reaching test under four conditions with different carrying loads, i.e. 0% Body Weight (BW), 5%BW, 10%BW and 15%BW. The subjects were tested under the barefoot condition. The order of carrying load was assigned according to a balanced Latin square method. The effects of load carriage weights, gender and movement direction were analyzed using a mixed repeated measure analysis of variance (RANOVA) with gender as the between-subject factor, and movement direction and carrying load as the within-subject factors. The level of significance was set at p=0.05.

In the second phase of the study, subjects were tested under different conditions of heel heights (i.e. flat shoes, 2cm high heel shoes and 5cm high heel shoes), and carrying loads (0%, 5%, 10% and 15% BW). There were totally 12 testing conditions in this part of the study. Latin square method was adopted to control the sequence of the testing conditions. Repeated measure analysis of variance (RANOVA) was used to analyze the effect of movement direction, load carriage and heel height as well as their interactions. The level of significance was set at p=0.05.

3.2 Carrying Load and High-heeled Shoes

As a conventional carrying object would not allow the subject's back to be exposed for measurement, a special two-strap suspension metal frame was adopted as the carrying load in this study (Figure 3.1). The middle part of the metal frame was removed to allow participant's back exposed for motion analysis. Additional dead weights were attached to the frame symmetrically about the midline of the mental frame so that the total weight of the frame could be adjusted to be equivalent to 5%, 10% and 15% of the participant's body weight (Chow et al., 2007). Foams were added as interface to enhance comfort.

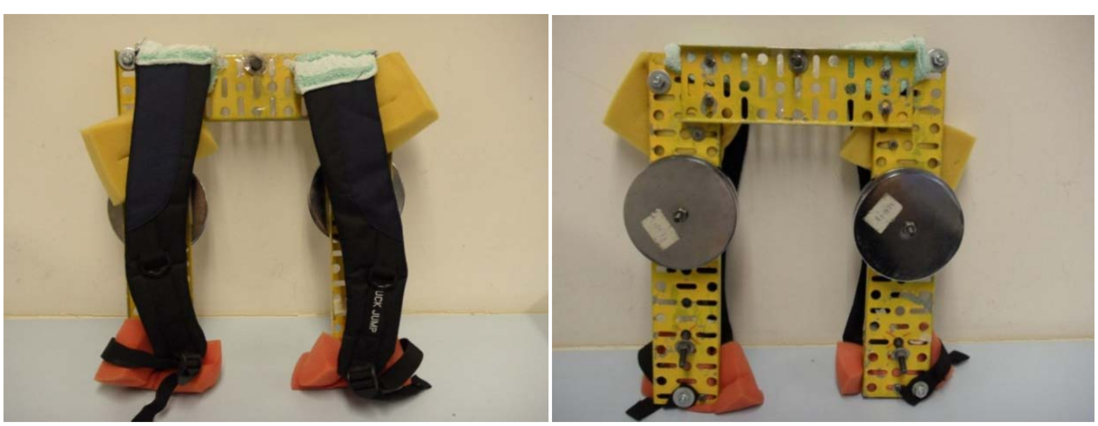


Figure 3.1 The special load carriage used in this study.

Common high-heeled shoes were bought from the market, three different heel-heights were chosen, i.e. flat, 2cm and 5cm.

Figure 3.2 High-heeled shoes used (from left to right: 5cm heel, 2cm heel and flat shoes)

3.3 **Subjects**

This study was approved by the Human Ethics Sub Committee of the Department of Health Technology and Informatics, The Hong Kong Polytechnic University. All the participations were recruited from the university. A written invitation letter together with a project information sheet (Appendix 1) was provided to each participant and a written informed consent form (Appendix 1) was obtained prior to the experiment.

Totally, 8 males and 8 females participated in Phase I, and another 12 females joined the Phase II of the study. Any participant with known musculoskeletal or neurological disorder, or history of shoulder or spinal disorders in the previous 12 months was excluded from the study.

3.4 **Experimental Procedure**

Prior to the experiment, subject's anthropometric data including age, body weight and body height were collected. An electronic bathroom scale (Tanita, HD-313, Tanita Corporation Tokyo, Japan) was used to measure the subject's body weight. Body weight was used to calculate the required weight of load carriage. A motion analysis system (VICON Nexus, Oxford Metrics, Oxford, UK) consisted of eight cameras was used to capture three-dimensional coordinates of reflective markers attached to the subject.

The subjects were advised for not to participate in any intensive physical activities on the day before the experiment to avoid possible fatigue. Prior to the experiment, various anatomical landmarks of the participants were identified. These included a protruded marker which was attached to be perpendicular to back surface proximal to the 1st lumbar vertebrae (L1), 3cm bilaterally at the two sides of L1, bilateral anterior superior iliac spines, bilateral posterior superior iliac spines and right hand, greater trochanter, knee (Figure 3.3). Spherical retro-reflective markers were affixed to the participants' skin surface proximal to the anatomical landmarks using double-sided adhesive tape. The markers were used to identify the positions and orientations of the lumbar spine and pelvis. The markers were attached to the participants when they adopted a semi-fixed position to minimize the effect of skin tension (Chow et al., 2007). Each participant was allowed to familiarize with the testing frame with no added weight and shoulder straps were adjusted, if necessary, for the best comfort.

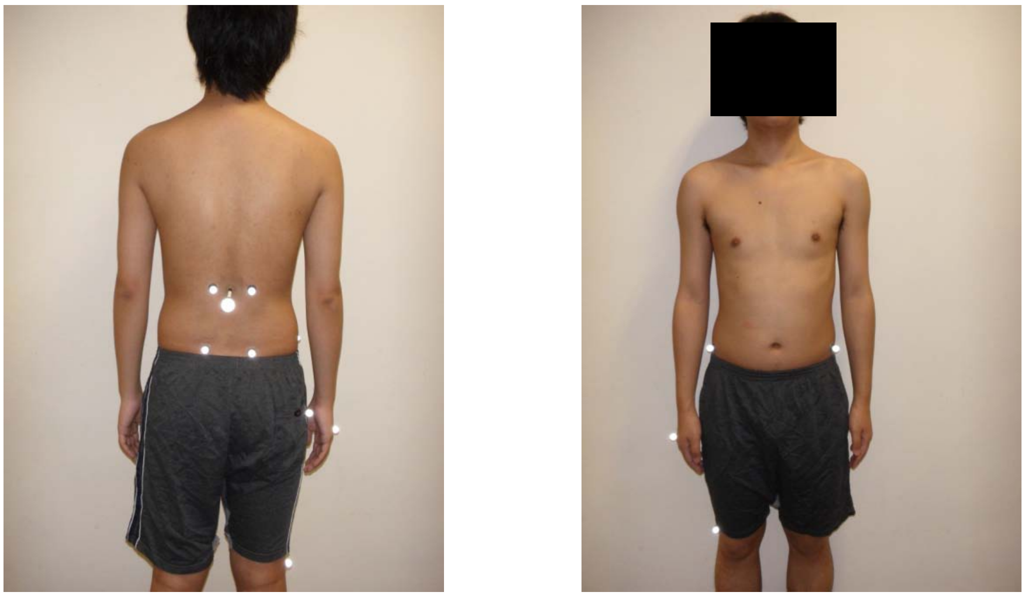


Figure 3.3 Marker placement (Left: Back view; Right: Front View)

A reaching test was conducted to quantify the movement coordination of the subject's spine in performing a functional reaching task (Silfies et al., 2009). Neuromuscular control of the spine and pelvis was investigated using the dynamical systems theory (DST) approach proposed by Silfies et al. (2009). Initially, the subject was asked to stand upright with feet at shoulder width, shoulders in 90° flexion and fully extended elbow. The subject was then instructed to perform a forward-reaching task. The subject was asked to perform three consecutive and continuous movements which consisted of forward reaching to a midline target located at shoulder height and returning to the upright standing posture. The subject was instructed to reach forward using his trunk and hips, which simulated the motion in reaching over a counter into a cupboard, touch a stationary target and immediately return to upright standing. The target distance for each subject was standardized to 50% of individual's functional reach determined under different testing conditions. Each subject's functional reach distance was determined by a Functional Reach Test (Duncan et al., 1990) prior to the experiment. The subject was required to stand with arm outstretched at shoulder height and reach as far forward as he could without taking a step (Figure 3.4). Standardizing subject's reaching distance to 50% of individual's functional reach was adopted to assess the control of trunk motion in mid-range where the neuromuscular system is primarily responsible for trunk dynamic stability (Figure 3.5). The subject would have 6s (3s forward, 3s back) to perform each repetition under the rhythm of a metronome. The movement trajectory of the lumbar-pelvic spine was captured by the motion analysis system. The subject was given three warm-up trials prior to each test. For the testing trial failed to touch the target or not able to maintain the standardized speed of motion, the trial would be repeated until 6 successful trials were acquired.

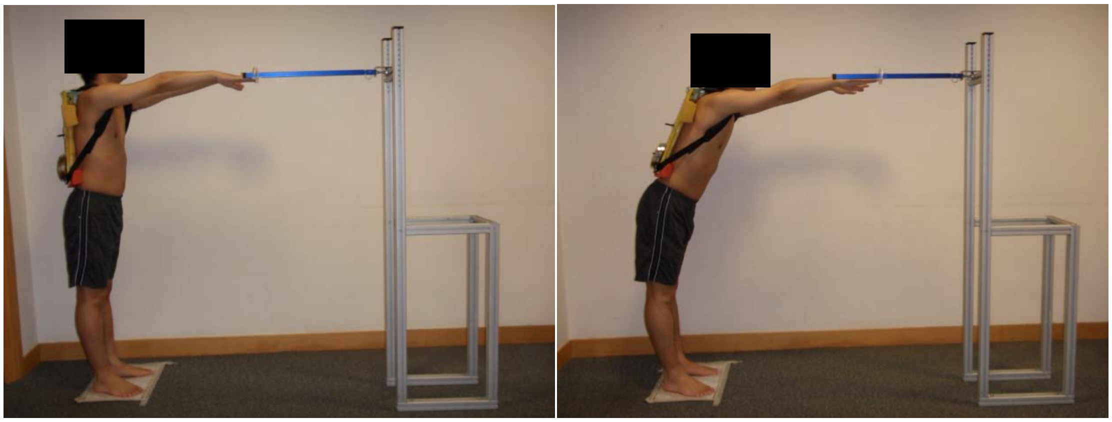


Figure 3.4 Functional reaching test. The subject was instructed to reach as far as possible without taking a step, and the maximum reaching distance was recorded.

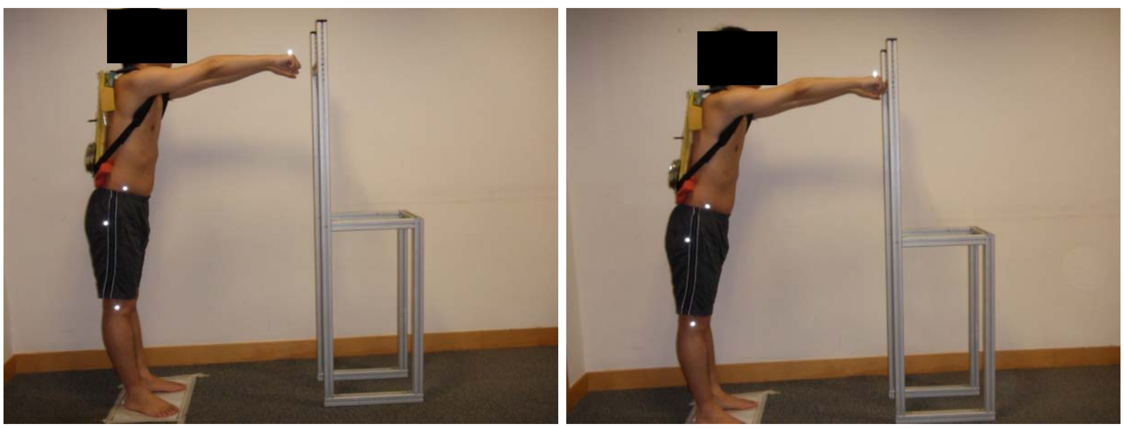


Figure 3.5 Mid-way of the functional reach test with 50% of the maximum displacement. In reaching test, the reaching distance was standardized to 50% of the maximum reaching displacement.

3.5 **Data Processing and Analysis**

Coordinates of the markers attached to subject’s lumbar spine and pelvis (Figure 3.6) were sampled at 50Hz by the motion analysis system.

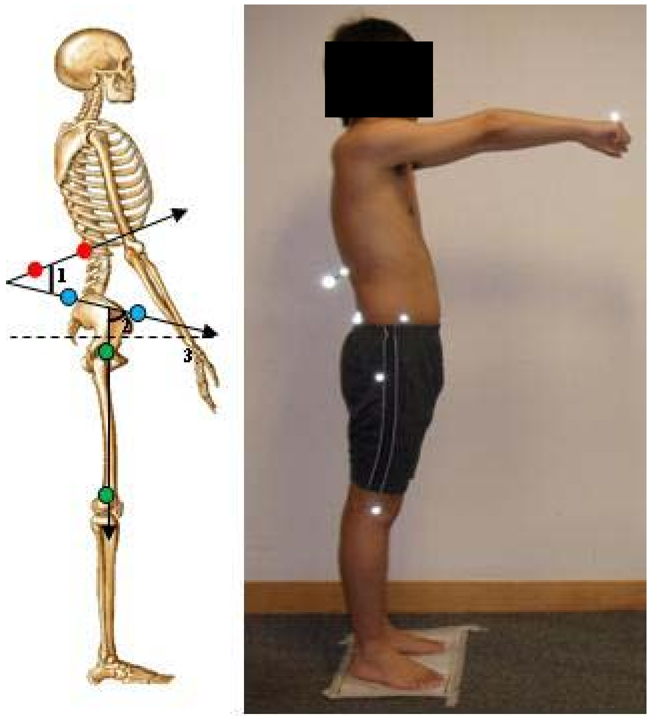


Figure 3.6 Marker placement (Lateral View) and angle calculation. Angle θ_1 was calculated to represent the posture and movement of lumbar spine. Angle θ_2 was calculated to represent the movement of pelvic relative to the thigh. Angle θ_3 was pelvic tilting relative to horizontal to represent the posture of pelvic.

A low pass filter (Matlab 2006b, Mathworks, Inc. USA) was used to filter the raw data and remove the noise (order 3, cut-off frequency 6Hz). The cut-off frequency was calculated by the residual method suggested by (Winter, 2005). Initial upright posture was recorded for 3s as the reference starting position. The standard deviations of initial upright posture between the six trials were calculated as repositioning error. Angular displacements and velocities of the subject's lumbar spine and pelvis were calculated (Silfies et al., 2009). The data of each repetition was time normalized to 40 data points, 20 for forward motion and 20 for backward motion. In order to quantify the movement coordination, a phase portrait was generated for each body segment (lumbar and pelvis) by plotting the angular displacement against the angular velocity of the segment. The resulting phase trajectory was used to calculate the phase angle of each data point throughout the entire motion using the following equation.

$$\psi = \tan^{-1}\left(\frac{\text{velocity}}{\text{displacement}}\right) \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots (1)$$

A continuous relative phase (CRP) curve was derived from the difference between the phase angles of pelvis and lumbar spine. This CRP curve denoted the coordination between the actions of the two interacting segments during a specific time period.

$$\varphi = \left|\psi_{\text{lumbar}} - \psi_{\text{pevlis}}\right| \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots (2)$$

To test the differences between CRP curves, the curves were quantified by two additional parameters which were derived using the ensemble curves method proposed by Stergiou et al. (2001). The first parameter termed mean absolute relative phase (MARP) was calculated as the average of the relative phase values over the CRP curve.

$$\text{MARP} = \sum_{i=1}^{20} \frac{|\varphi_{\text{relativephase}}|_i}{20} \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots (3)$$

A low MARP value would indicate a more in-phase relationship or segments moved in a similar manner while a high MARP value would indicate a more out-of-phase relationship or segments moved in opposite directions.

The second parameter named deviation phase (DP) was calculated by averaging the standard deviations of the ensemble CRP curve points.

$$DP = \sum_{i=1}^{20} \frac{SD_i}{20} \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots (4)$$

Functionally, a low DP value would indicate a more stable (i.e. less variable) organization of the neuromuscular system and a high DP value would indicate an instability in the organization of the neuromuscular system.

Besides, lumbar movement ratio was also determined in terms of the absolute maximum movements of the lumbar spine to the sum of lumbar and pelvis.

The means and standard deviations of the initial upright posture, functional reaching distance, MARP, DP values，lumbar movement ratios and demographic data were determined using PASW Statistics 18.0 software (SPSS Inc. Chicago, IL, USA). These parameters were analyzed using repeated measure analysis of variance (RANOVA). The level of significance level was set at p=0.05.

Chapter 4 Results

4.1 Details of Participants

For the barefoot experiment, totally eight males and eight females were recruited (Phase I). In the high-heeled shoes part (Phase II), another twelve female students participated in the study. Demographic data of the subjects are summarized in Table 4.1. Although some subjects reported that the 15%BW backpack was very heavy, they could tolerate it and none of them complained back discomfort or back pain during the test.

Table 4.1 Participants' information

	Phase I Overall	Phase I		Phase II
		Male	Female	
Mean (SD) age (year)	**26.2 (2.3)**	**27(2.4)**	**25.4 (2.0)**	**20.9 (1.3)**
Mean (SD) body height (cm)	**166.8 (8.2)**	**173.4 (2.9)**	**160.3 (8.2)**	**158.4 (4.0)**
Mean (SD) body weight (kg)	**62.1 (10.9)**	**67.8 (7.3)**	**56.3 (11.2)**	**50.9 (5.3)**
Number of subjects	**16**	**8**	**8**	**12**

The dependent variables, mean absolute relative phase (MARP) and deviation phase (DP), were analyzed by repeated measure ANOVA with mixed samples with load carriage weights (0%, 5%, 10%, 15%BW) and movement direction (forward and backward) as within-subject factors and gender (male and female) as between-subject factor using PASW Statistics 18.0 s oftware (SPSS Inc. Chicago, IL, USA). The additional variables, functional reaching distance, initial upright posture and lumbar movement ratio were analyzed with only loading condition as the within-subject factor. The data met the assumption of ANOVA. The level of significance was set at p=0.05 throughout and pos-hoc comparisons were made using LSD criterion. The no l oad condition was used as the baseline.

Similarly, in Phase II of this study, heel height was another within-subject factor besides load carriage and movement direction. No gender effect was investigated in this session as all of the participants were female.

4.2 **Effects of Load Carriage (Phase I)**

4.2.1 **Initial Upright Posture**

The initial upright postures of lumbar and pelvis under each load condition were determined. The interactions of gender*loading and main effects of gender and loading were not statistically significant with p>0.05 for both lumbar and pelvis (Figure 4.1). It was observed that initial lumbar lordosis of males was consistently greater than that of females, and the pelvis titling of males was more horizontal compared to females who tilted forward.

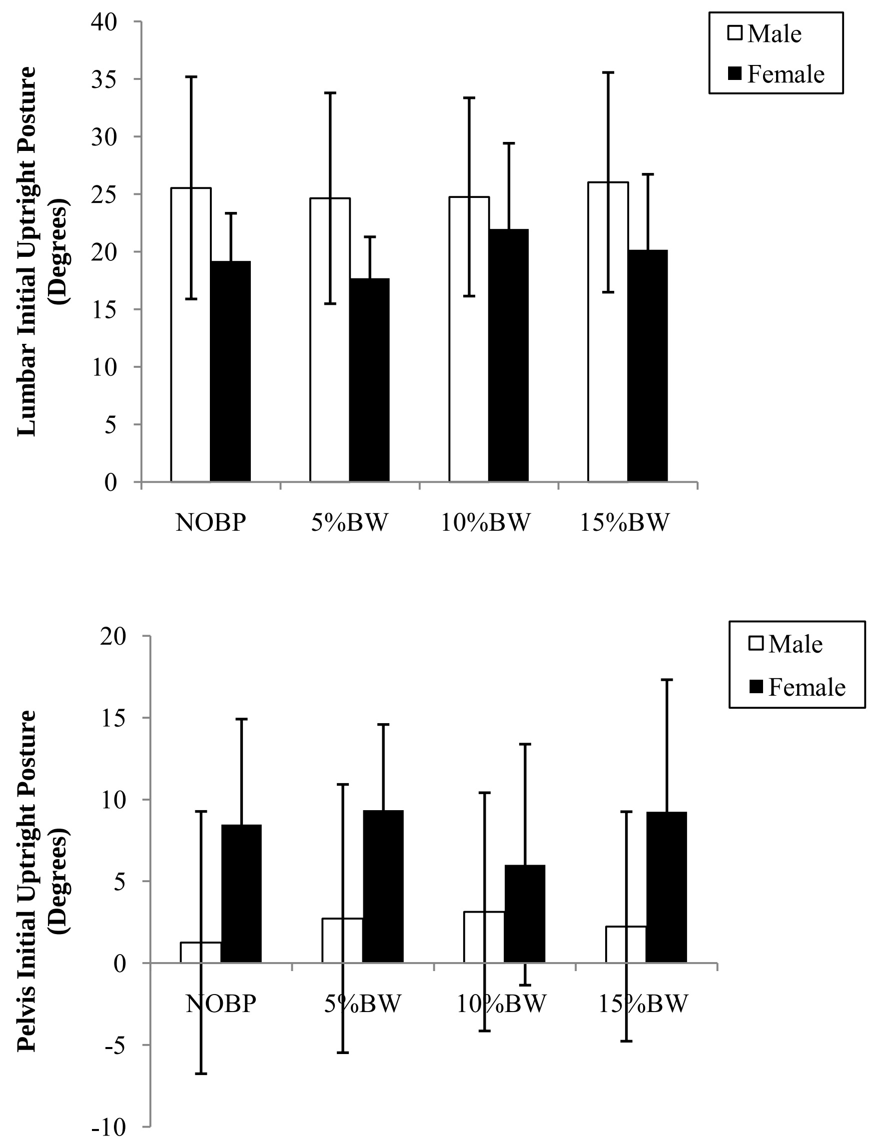
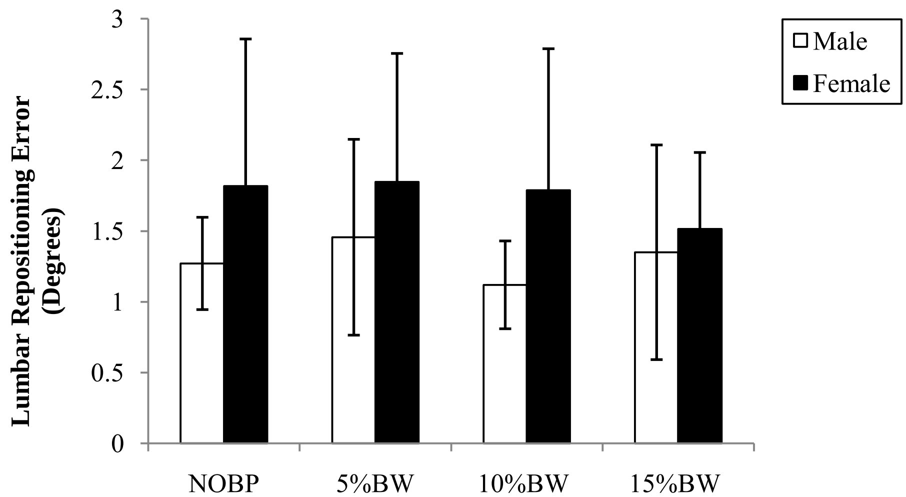


Figure 4.1 Mean values and standard deviations of initial upright posture for all subjects, Lumbar lordosis (Above) and pelvic tilting (Below) respectively.

4.2.2 **Repositioning Error**

The repositioning errors of lumbar and pelvis under each load condition were determined. The interactions of gender*loading and main effects of gender and loading were not statistically significant with $p>0.05$ for both lumbar and pelvis (Figure 4.2). It was observed that repositioning error of males was consistently smaller than that of females under each condition for lumbar and pelvis.



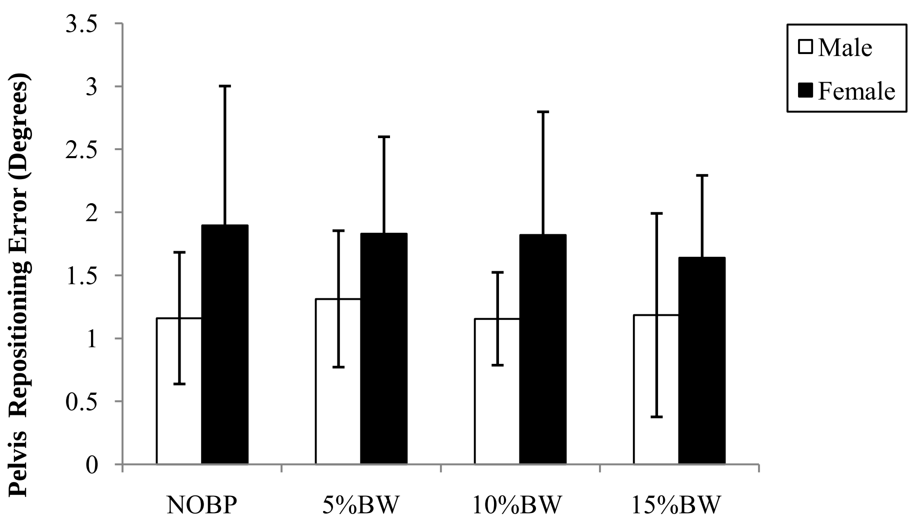


Figure 4.2 Mean values and standard deviations repositioning error for all subjects, Lumbar (Above) and Pelvis (Below) respectively.

4.2.3 **Functional Reaching Distance**

The mean reaching distance under each load condition was determined. The interactions between gender and loading condition and the main effect of gender was not statistically significant with p>0.05 (Figure 4.3). The main effect for loading weight on r eaching distance was significant with p<0.001. There was significant reduction in reaching distance during the 5%BW, 10%BW and 15%BW load conditions compared to the no load condition (Figure 4.3). However, a general trend of decreasing reaching distance was found with increasing load carriage (Figure 4.3). It was also observed males could reach further than females under each condition consistently.

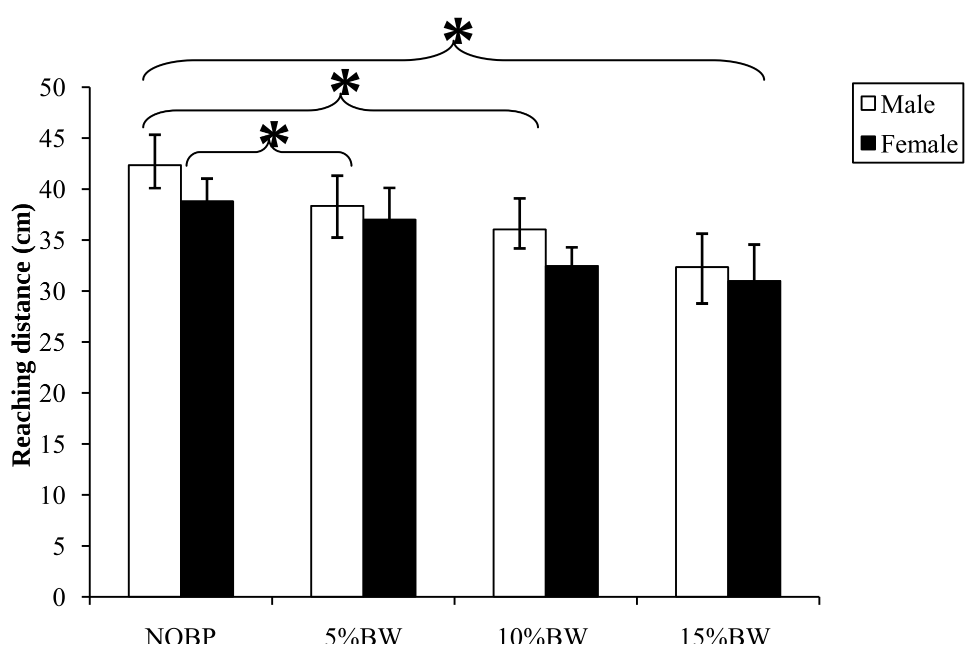


Figure 4.3 Mean values and standard deviations of maximum reaching distance for all subjects. Decreased reaching distance was observed with increasing loading (* indicated significantly different from no load condition).

4.2.4 **Mean Absolute Relative Phase**

Movement coordination between lumbar spine and pelvis was assessed using the parameter of mean absolute relative phase (MARP). The mean MARP value under each condition was determined. The interactions among the three main factors were not statistically significant with $p>0.05$, while the main effects for loading, gender and movement direction were all significant with $p<0.05$ (Figure 4.4). It was found that MARP during backward motion was significantly larger than that of forward motion. Additionally, MARP for males was larger than that of females significantly. It was also observed that the MARP had a slight increase for 5%BW loading condition, however the change was not statistically significant compared to the no load condition. There was a significant change between 10%BW, 15%BW compared to the no load condition (Figure 4.4). Moreover, a general trend of increasing MARP value was found with increasing load carriage.

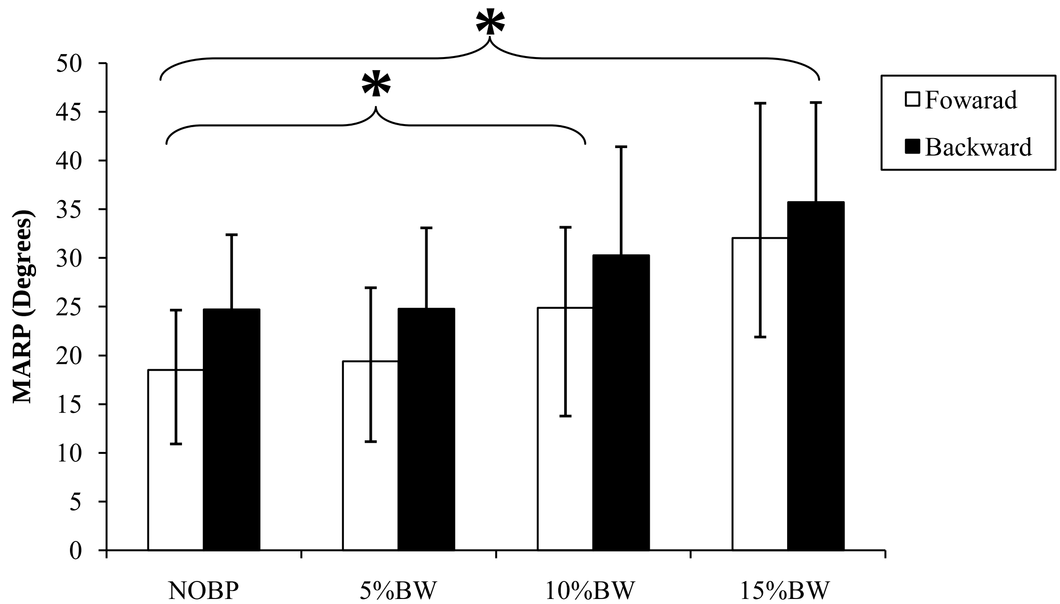


Figure 4.4 Mean values and standard deviations of mean absolute relative phase (MARP) for all subjects. Increased MARP value was observed with increasing loading (* indicated significantly different from no load condition).

4.2.5 **Deviation Phase**

Stability or variability of each individual's behavior was assessed by the parameter of deviation phase (DP). The mean DP value under each condition was determined. The interactions of loading*gender, gender*movement direction, loading*gender*movement direction and main effects for gender and movement direction were not statistically significant with $p>0.05$. However, the interaction between loading and movement direction were significant statistically. Thus, 2-way Repeated Measures ANOVA with mixed samples was applied to investigate the effects of loading weight and gender for each movement direction. It was found that the main effects of gender on the DP values were not statistically significant for both forward and backward movement directions with p>0.05, while the effect of carrying load was significant for both directions (p<0.05). For the forward motion, there was a significant change only during the 10%BW and 15%BW load condition compared to the no load condition. The difference was also significant only under 15% BW load condition for backward motion. A general increasing trend was also found with increasing load carriage (Figure 4.5).

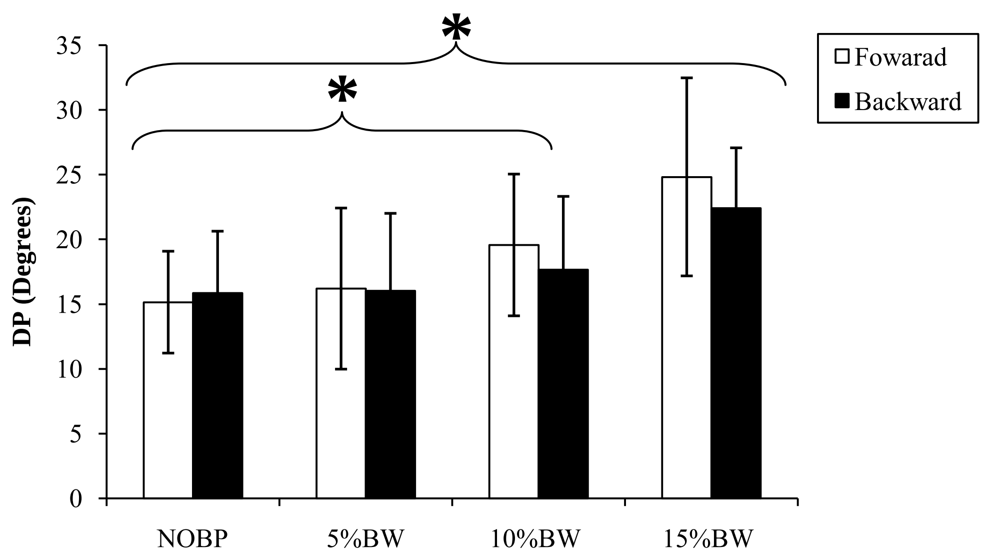


Figure 4.5 Mean values and standard deviations of deviation phase (DP) for all subjects. Trend of increased DP value was observed with increasing loading (* indicated significantly different from no load condition).

4.2.6 **Lumbar Movement Ratio**

The mean movement ratio of lumbar under each condition was determined. The interactions between the two main factors gender*loading and the main effects of gender were not significant with p>0.05, while the main effect of loading was significant (Figure 4.6). The ratios under 10%BW and 15%BW conditions were significantly larger than that of the no l oad condition. A general increasing trend was found with the increasing weight of loading, though there was no significant difference between 5%BW and no load condition. It was also observed the ratio of males was consistently slightly larger than that of females.

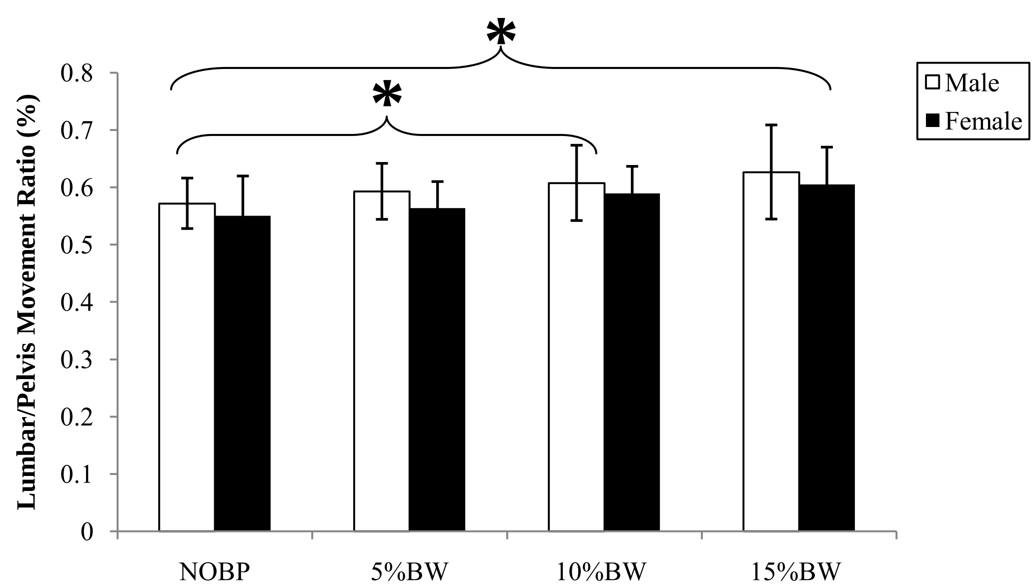


Figure 4.6 Mean values and standard deviations of lumbar movement ratio for all subjects. Trend of increased ratio value was observed with increasing loading (* indicated significantly different from no load condition).

4.3 **Combined Effects of Load Carriage and High-heeled Shoes (Phase II)**

4.3.1 **Initial Upright Posture**

The mean initial upright postures of lumbar and pelvis under each load condition were determined. For lumbar lordosis, the interactions between loading and heel height and the main effect of loading were not statistically significant with $p>0.05$, while the main effect of heel-height was significant (Figure 4.7). The lumbar lordosis was significantly decreased with 5cm high-heeled shoes compared to the flat shoes condition.

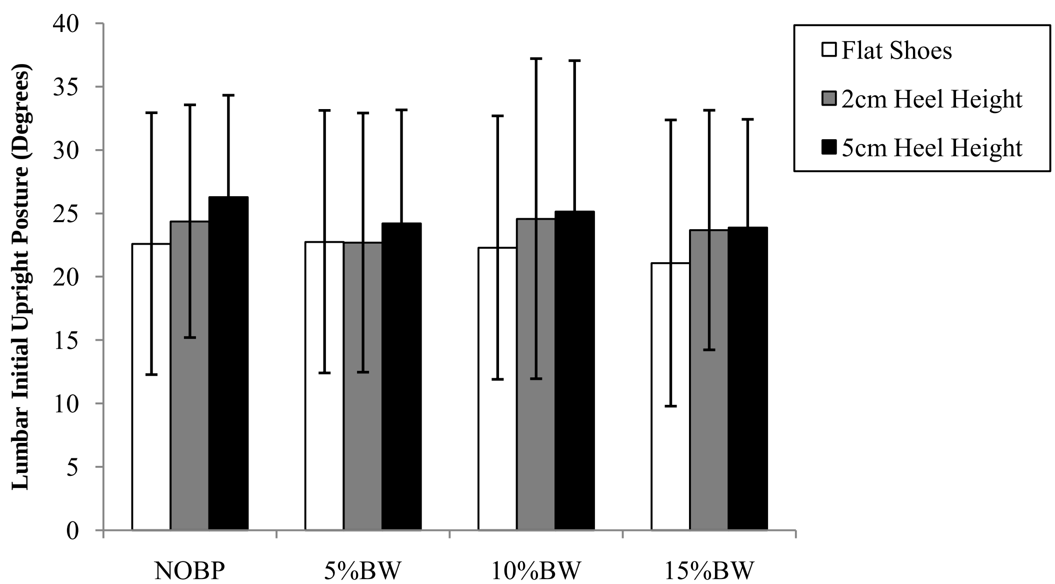


Figure 4.7 Mean values and standard deviations of lumbar initial upright posture for all subjects.

For pelvis tilting, the interactions between loading and heel height and the main effect of loading were not statistically significant with p>0.05, while the main effect of heel-height was significant (Figure 4.8). The pelvis was tilted backward significantly wearing 2cm and 5cm high-heeled shoes compared to the flat shoes condition.

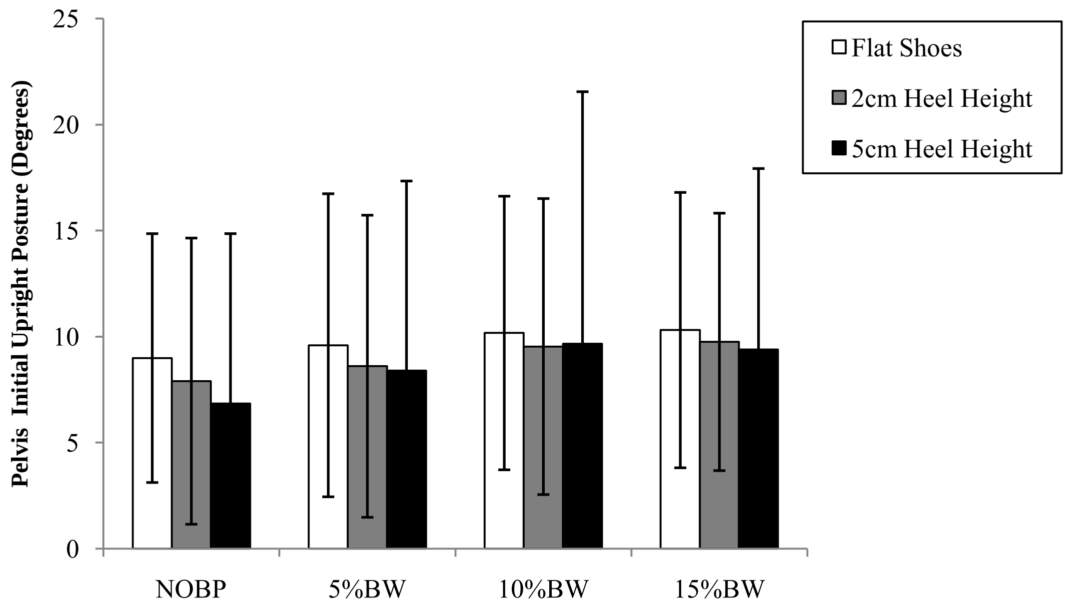


Figure 4.8 Mean values and standard deviations of pelvis initial upright posture for all subjects.

4.3.2 **Repositioning Error**

The mean repositioning errors of lumbar and pelvis under each load condition were determined. For the lumbar repositioning error, the interactions between loading and heel height and the main effect of heel height were not statistically significant with p>0.05, while the main effect of loading was significant (Figure 4.9). The repositioning errors of lumbar spine were significantly increased under 5%BW, 10%BW and 15%BW conditions compared to the no load condition.

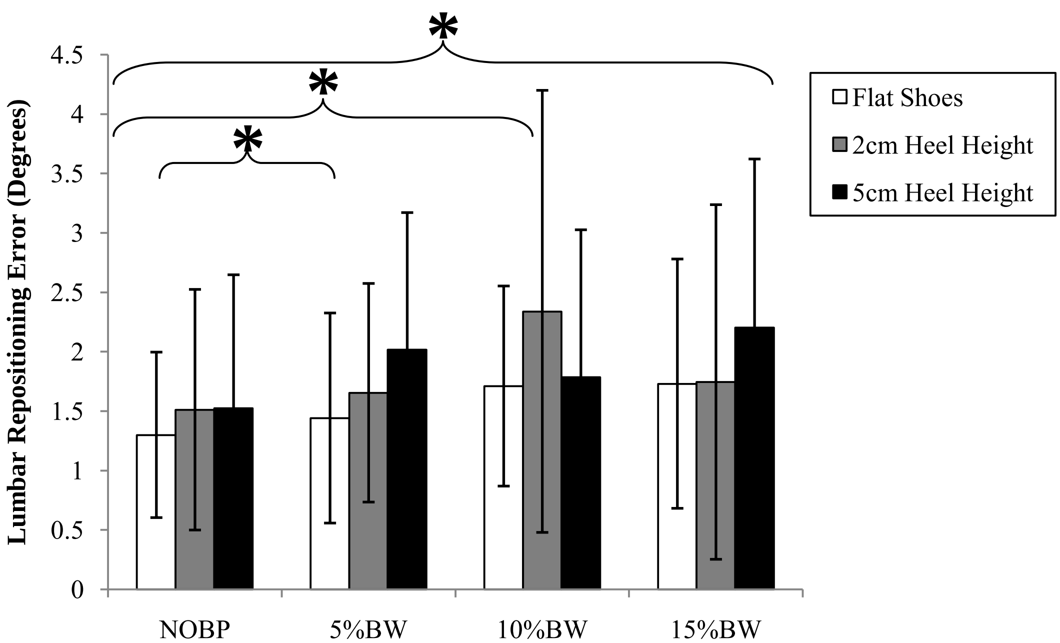


Figure 4.9 Mean values and standard deviations of repositioning error of lumbar spine for all subjects (* indicated significantly different from no load condition).

For the pelvis repositioning error, the interactions between loading and heel height and the main effect of loading were not statistically significant with $p>0.05$, while the main effect of heel-height was significant (Figure 4.10). The repositioning error of pelvis was significantly increased with 5cm high-heeled shoes compared to the flat shoes condition.

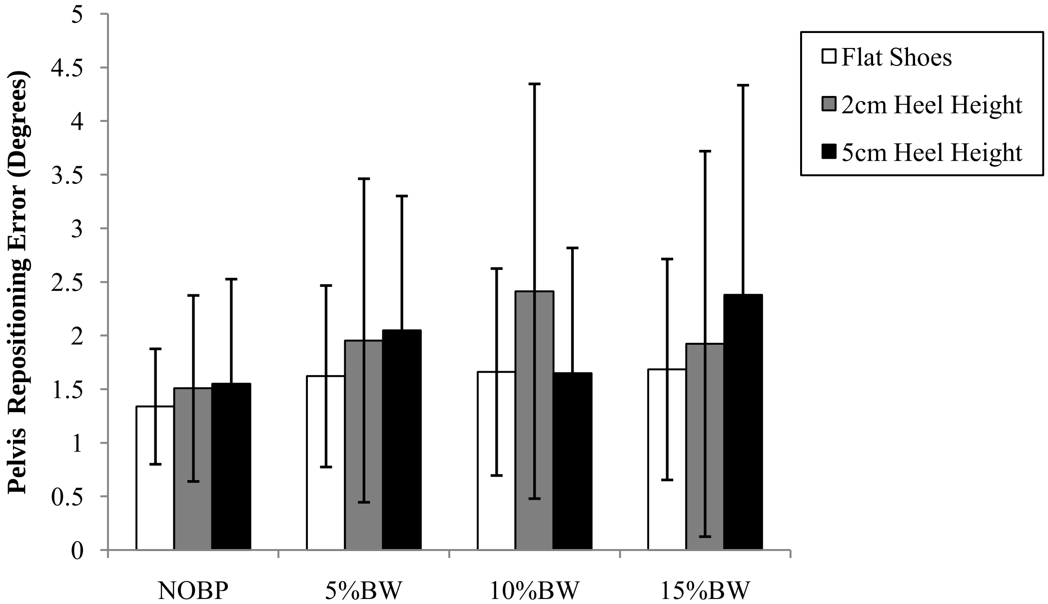


Figure 4.10 Mean values and standard deviations of repositioning error of pelvis for all subjects.

4.3.3 Functional Reaching Distance

The mean functional reaching distances under various conditions were determined and the interactions between the two main factors, loading and heel height was not statistically significant with p>0.05 (Figure 4.11). The reaching distance under no load and flat shoes condition was about 35cm. It was decreasing gradually to around 25cm at the 15%BW and 5cm high-heeled shoes condition (Figure 4.11). The main effects for load carriage and high-heeled shoes both have significant effects on the reaching distance with p<0.001. The reaching distances under 5%BW, 10%BW and 15%BW were significantly shorter than that of the no load condition. It was also observed that subjects reached shorter significantly with 2cm or 5cm high-heeled shoes compared to the flat shoes condition.

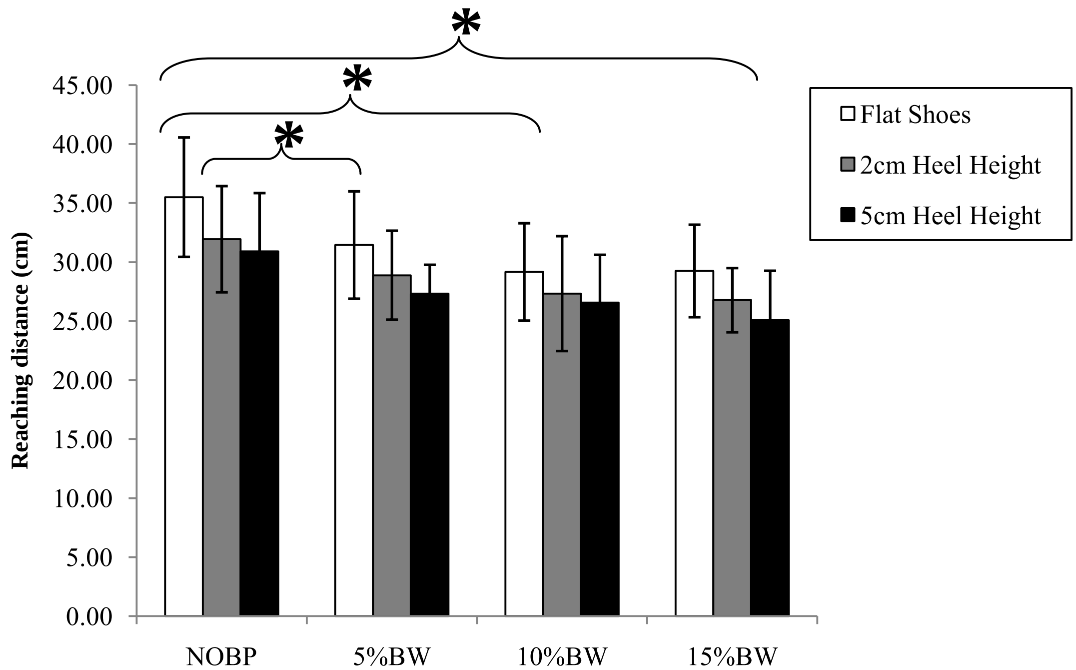


Figure 4.11 Mean values and standard deviations of maximum reaching distance. Decreased reaching distance was observed with increasing loading and heel height (* indicated significantly different from no load condition).

4.3.4 Mean Absolute Relative Phase

The mean MARPs under various conditions were determined. The interactions of loading*movement direction, loading*heel height and loading*heel height*movement direction were not statistically significant with $p>0.05$, while the main effects of loading, heel height and movement direction were all significant. The MARP value under 15%BW loading condition was significantly larger than that of the no load condition. The MARP values wearing 2cm and 5cm high-heeled shoes conditions were significantly smaller than that of the flat shoes condition Moreover it was also found that MARP during forward motion was significantly smaller than that during the backward motion.

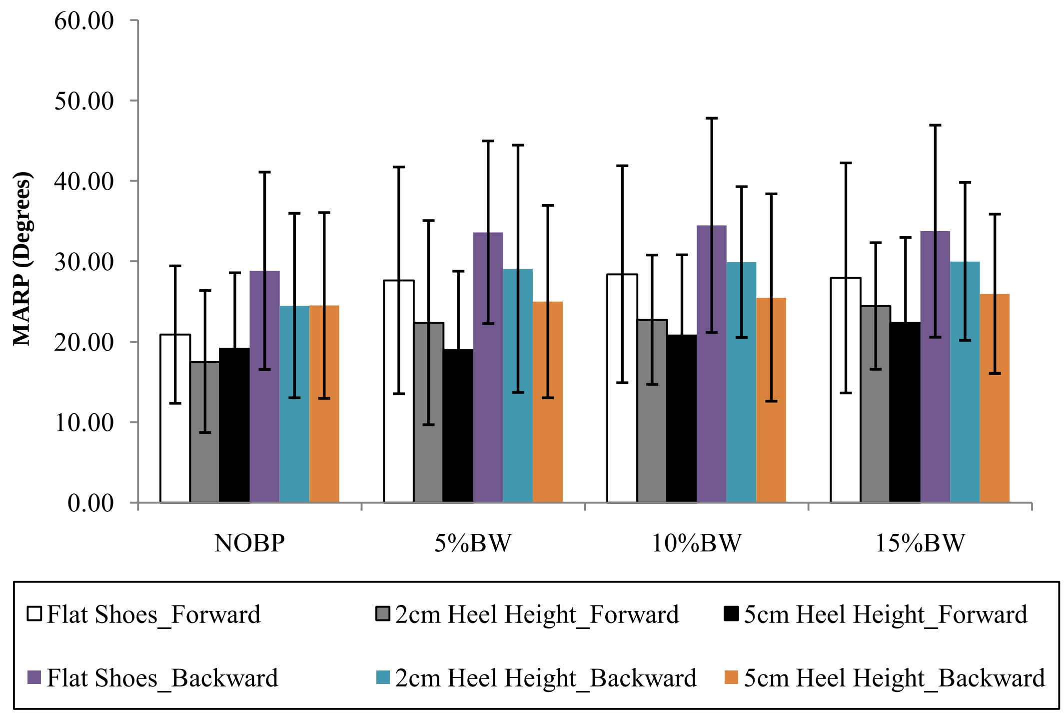


Figure 4.12 Mean values and standard deviations of mean absolute relative phase (MARP) for all subjects in both forward and backward direction. Decreased MARP value was observed with increasing heel height.

4.3.5 **Deviation Phase**

The mean DP value under each condition was determined. The main effect of movement direction and the interactions of loading*movement direction, loading*heel height and loading*heel height*movement direction were not statistically significant with p>0.05, while the main effect of loading and heel height was significant with $p<0.05$ (Figure 4.13).

The DP value under 15%BW loading condition was significantly larger than that of the no load condition. The DP values wearing 2cm and 5cm high-heeled shoes conditions were significantly smaller than that of the flat shoes condition. Under most conditions, it was also observed that DP during forward motion was smaller than that during the backward motion though the difference was not statistically significant.

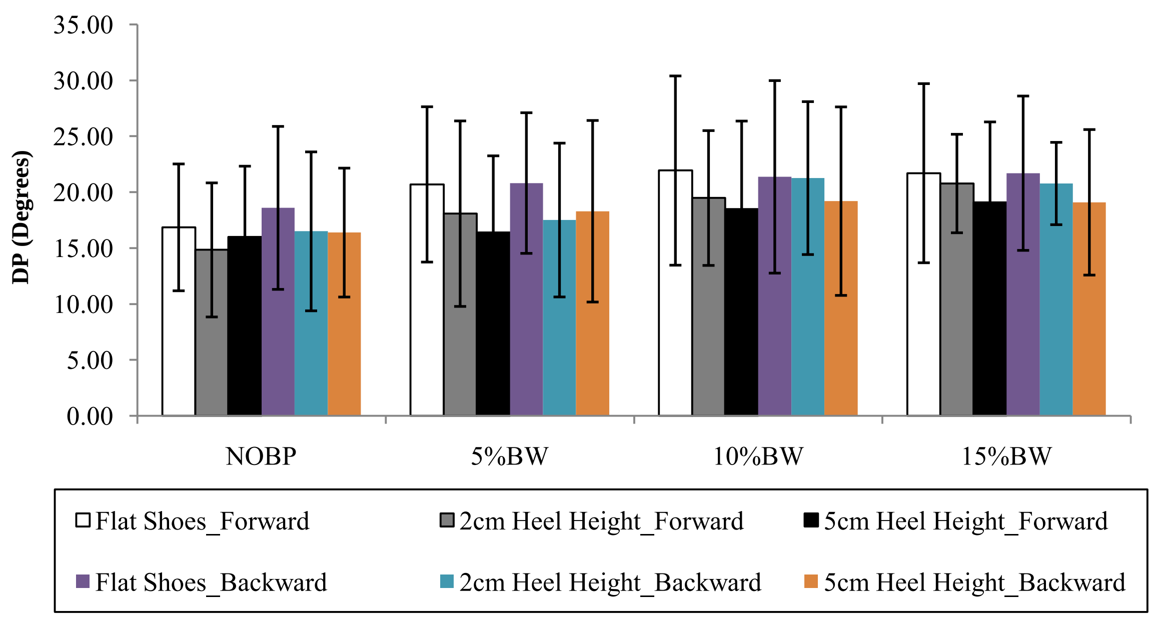


Figure 4.13 Mean values and standard deviations of deviation phase (DP) for all subjects in both forward and backward direction. Decreased DP value was observed with increasing heel height.

4.3.6 **Lumbar Movement Ratio**

The movement ratio of lumbar under each condition was determined. The interactions between the two main factors loading*heel height and the main effects of loading were not significant with $p>0.05$ statistically, while the main effect of heel height was significant with $p<0.001$ (Figure 4.14). This ratio under 2cm high-heeled shoes and 5cm high-heeled shoes conditions were significantly smaller than that of the no load condition.

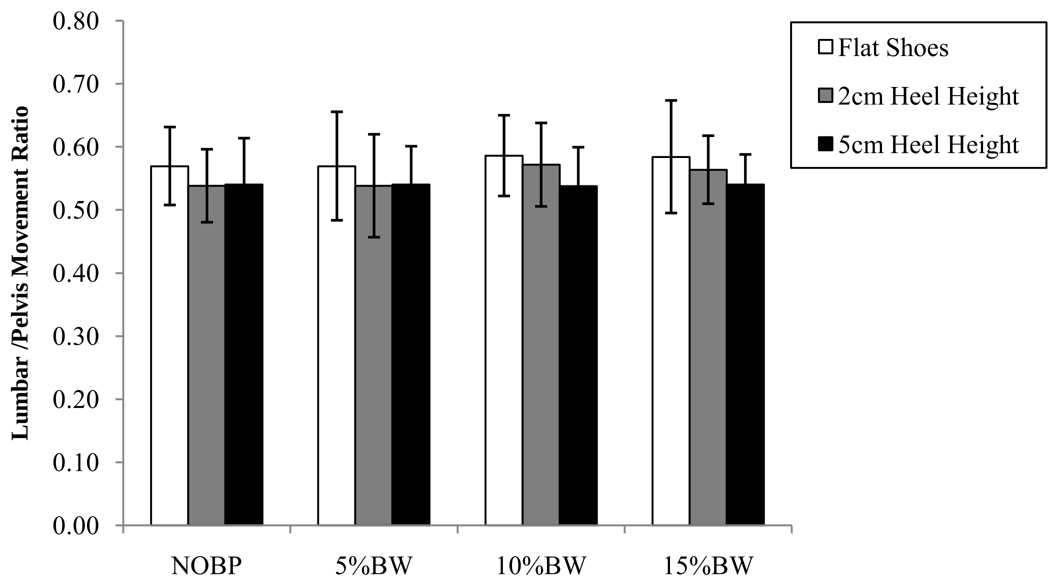


Figure 4.14 Mean values and standard deviations of lumbar movement ratio for all subjects. Trend of decreased ratio value was observed with increasing heel height.

Chapter 5 Discussions

The present study provides useful information on the normal kinematic patterns of the spine and pelvis. The measurement technique was found to provide repeatable data. Both quasi-static and dynamic performance assessments were conducted to investigate the effects of load carriage and high-heeled shoes on spinal motor control.

5.1 Quasi-static Assessment

Posture and repositioning error represented the performance under quasi-static situation. From the results, lumbar flattening and backward pelvic tilting were observed in wearing high-heeled shoes. This finding was consistent with those reported by Opila et al. (1988) and Bendix et al. (1984) who investigated postural alignment in barefoot and high-heeled stance. Repositioning consistency depends on i ntegrated sensory information from visual, vestibular and proprioceptive inputs (Magnusson et al., 2008). The sensory input was standardized as far as possible by providing a constant target for the participants to fix their gaze on and use of the standardized reaching test posture. In the current study, we further demonstrated that repositioning consistency at the lumbar spine and pelvis was compromised during load carriage and high-heeled shoes, respectively. The spine was subject to greater variations in stress and strain because of the increased variability of spinal posture. It was more difficult for an individual to maintain the natural spinal posture in demanding conditions. In addition, a number of studies have reported a significant decreased repositioning performance in the low back pain group (Brumagne, Lysens, & Spaepen, 1999; Kara, Genc, Yildirim, & Ilcin, 2011; Newcomer, Laskowski, Yu, Johnson, & An, 2000; O'Sullivan et al., 2003). We hypothesized that the load carriages and high-heeled shoes might result in a high demand by affecting the positioning sense of the spine and might be potential risk factors on spine musculoskeletal disorder. The clinical evaluation of the effects of load carriage and high-heeled shoes in relation to the chance of increase in back problems should be further investigated.

5.2 Dynamic Assessment

Reaching test was employed as the standardized movement task to investigate the dynamic effects in this study. Besides forward reaching, lateral bending, forward and backward bending and twisting have also been used by other researchers (Lee & Wong, 2002; Wong & Lee, 2004). One of the reasons why reaching test was chosen was that specific strategy of trunk components in performing reaching motion in isolation from limb motion could be examined. Moreover, we could measure the functional reaching distance that has been a co mmonly used clinical parameter for evaluating balance control and the margin of stability (Duncan et al., 1990). Functional reaching distance is a key and direct measure of balance control. To eliminate the subjective effects on the results of reaching distance, standard instructions were given to the participants. In addition, three trials were conducted and the mean value of reaching distance was determined as final functional reaching distance. Additionally, forward reaching is a common daily movement to all participants, any learning effect on the results could be minimized. In this study, the reaching distance for testing lumbar-pelvis coordination was standardized to 50% of the functional reach distance and the pace of motion was standardized by a metronome. These were used to enhance the repeatability of the measurements. Reaching distance, mean absolute relative phase, deviation phase and lumbar movement ratio were determined to evaluate the dynamic performance under each condition.

From the results, reaching distance was decreased with both carrying load and high-heeled shoes, which is a sign of balance control deficit (Duncan et al., 1990). Mean absolute relative phase (MARP) has been used to quantify whether the interacting segments display an in-phase or out-of-phase pattern during the movement. Deviation phase (DP) of the relative phase between two interacting segments has been used to determine variation in the organization of the neuromuscular system which represents the stability of movement (Stergiou et al., 2001). These parameters have been widely used to investigate the gait patterns in previous studies (Stergiou et al., 2001; Thelen & Ulrich, 1991; Winstein & Garfinkel, 1989). In this study, these two parameters were used to represent the movement relationship between lumbar and pelvis during a reaching test, and both of them were found to be affected by load carriage and high-heeled shoes in this study. It was found that the effects of load carriage and high-heeled shoes on MARP and DP were opposite. The movement was more in phase and stable in wearing high-heeled shoes, while it was more out-of-phase and vary during load carriage. These changes may be related to the posture changes. However, further studies were required to examine this hypothesis. It was also interesting to observe that MARP in forward motion was significantly smaller than that in backward motion. The DP in forward motion was consistently larger than that in backward motion though the difference is not significant. This indicated that the movement coordination between lumbar and pelvic was more in-phase during forward reaching but with larger movement variability. This bilateral reaching task requires predominant contribution from back and hip extensors, in order to further understand the mechanism why there was difference between forward and backward motion, electromyographic (EMG) studies monitoring the changes of spinal muscles activity during reaching test may help.

There was evidence that back pain could alter the relationship between the movements between lumbar spine and hip in the sagittal plane (Esola et al., 1996; Porter & Wilkinson, 1997). It is therefore, important that clinical examination of back patients should include measurement of the movements of both the spine and pelvis. Altered movement patterns of the spine and hip may be a potential factor that contributes to the development of low back pain. For instance, Dolan & Adams (1993) showed that changes in spine and hip mobility would alter the bending stresses of spinal motion segment. On the other hand, in patients with low back pain, it may also be argued that altered movement patterns of the spine and pelvis might be the consequence of low back pain. It might also be a compensatory response to reduce pain or to protect tissues.

The ratios of the absolute maximum movements of the lumbar spine to the sum of lumbar and pelvis were determined. This described the relative contributions of the two joints at the end position. The mean ratios in both phases of the study were close to 0.5 (Figures 4.6 & 4.14). This suggested that the maximum ranges of motion for lumbar spine and pelvis were approximately equal. In phase I, it was found that this ratio was increased significantly with carrying load, which means that the contribution of the lumbar spine was larger during load carriage. This might be one of the potential risk factors of low back pain if the working load on the lumbar spine was increased during load carriage as the amount of flexion-extension motion was increased during load carriage. This could induce increased spinal loading as well as muscle overuse and fatigue. It was however, observed that high-heeled shoes could alleviate this ratio significantly. From the results of MARP, DP and movement ratio, though no interaction effect was found between loading and heel height, wearing high-heel shoes might help to alleviate the adverse effects due to load carriage.

5.3 Limitations and Future Work

Our results should be interpreted in light of the limitations of our work. First, analysis was limited to the sagittal plane, and trunk movement is three-dimensional. Moreover, we chose a task that minimized kinetic effect of limbs on trunk motion by having subjects with the arms extended and level. Second, we did not standardize postural alignment of the trunk and pelvis, but instead the subject adopted relaxed standing posture as the reference position. Third, since six trials were conducted for each condition, confounding effect of fatigue and learning might exist though enough rest has been given to the subjects between trials. Further study designs and concurrent EMG and kinematics may offer additional insight into the impaired control mechanisms during load carriage and high-heeled shoes. Furthermore, only posterior load carriage method was considered in this study. As shoulder bag, anterior or double pack methods are commonly used in our daily life, future studies could investigate the dynamical effects of different loading methods on spinal motor control.

In the currents study, the special metal frame was adopted as the carrying load in this study. Additional dead weights were attached to the frame symmetrically about the middle line of the metal frame (Figure 3.1). This open-channel carrying load is different from the daily used backpack for the reason that the subject’s back has to be exposed during the measurement by the motion analysis system. Also it was found that both spinal curvature and repositioning error were affected by backpack center of gravity (CG) level in previous study (Chow et al., 2010). However, the CG level was not controlled as a parameter in this study since it was varied only a few millimeters for different weights. Only the effects of carrying load weights were investigated in this study independent of CG level. This should be considered and improved in future study.

In order to investigate the effects of high-heeled shoes on s pinal motor control, all female subjects were recruited from the university who were not experienced high-heeled shoes wearers. However, it was reported that biomechanical accommodations to high-heeled shoes varied with age and experience in wearing high-heeled shoes (Opila-Correia, 1990). In this study, only the effects on inexperienced and young wearers were investigated, further studies including more subjects of different ages and experienced wearers could be conducted to give insights to this aspect.

In addition, the change in lumbar lordosis found in current study during high-heeled shoes was inconsistent with clinical findings of hyperlordosis in habitual wearers of high-heeled shoes who have increased lumbar lordosis and a forward tilted pelvis. This inconsistency might be explained in two ways: 1) in the laboratory, the subjects wore their shoes only for a short period of time before and during testing, this did not produce the posture variations associated with fatigue, or 2) the alignment of the lumbar-pelvic region during stance may not be the same as that during dynamic activities. To address this problem of disparity between spine observed laboratory results and clinical findings, two types of studies are indicated: 1) electromyographic studies evaluating postural muscle activity in different heel heights assessing changes in muscle activity with fatigue and 2) studies measuring the kinematics and kinetics of the lumbar and pelvis during gait in both low- and high-heeled shoes.

Despite these limitations, use of a dynamical systems approach to understand coordination patterns and organization of the neuromuscular system during load carriage and high-heeled shoes offers insights that may not be evident when using analysis of time series or discrete data alone. Our data indicated that inter-segmental coordination was altered and more variable during load carriage, similar results was also found in the patient with low back pain (Silfies et al., 2009) which suggested that load carriage might be a potential risk factor of low back pain. This tool could potentially be applied for determining the efficacy of trunk motor control exercises for treating patient with low back pain attributed to poor neuromuscular control.

Chapter 6 Conclusion

The current study applied dynamical systems theory to study the movement coordination and stability between lumbar spine and pelvis. In the first part of this study, it was found that load carriage could adversely affect spinal movement. The situation was getting worse with increased carrying load. The reaching distance was shorter during load carriage which indicated that balance control was compromised. The movement was more out of phase and more vary. There was no significant difference between male and female for the performance. The difference was significant under 10%BW and 15%BW load condition compared to the no load condition. A deeper understanding of the clinical implications of the reduced reaching distance, movement coordination and stability due to load carriage may provide insight whether this is related to the observed common back pain among users required heavy load carriage. It was suggested that clinical examination of back patients should include measurement of the movements of both the spine and pelvis.

It was hypothesized that the change of spinal motor control caused by load carriage was due to the change in body alignment and pelvic orientation during load carriage. In an attempt to alleviate the effects of load carriage, modification of pelvis orientation was proposed by wearing high-heeled shoes in Phase II of this study. It was demonstrated that the effects of load carriage and high-heeled shoes on spinal motor control were opposite. Thus, spinal motor control could be enhanced by wearing high-heeled shoes. However, there was a decrease in functional reaching distance when wearing high-heeled shoes. Further experiment is required to identify the optimal heel height for improving spinal motor control without increased risk of fall due to load carriage.

Appendices

Appendix 1: Invitation Letter, Information Sheet and Consent Form for Subject Recruitment

Part a:

Invitation to participate in project entitled

"Effects of Load Carriage and High-heeled Shoes on Spinal Motor Control"

Low back pain (LBP) is highly prevalent worldwide with about 80% of individuals suffering from LBP at some point in their life. There is evidence that recurrence of LBP is associated with impaired motor control of spine. Patients with LBP were shown to have delayed trunk muscle reflex response which has been suggested to be associated with increased risk of low back injuries and a cause of recurrent back pain. Backpack carriage has been identified as one of the mechanical risk factors for the increasing prevalence of low back pain in schoolchildren and subsequently this might increase the risk to experience chronic back pain in their later life. The results of previous studies showed that posterior load carriage had significantly effects on spinal curvature with reduction in spinal repositioning ability. Therefore, in order to gain a deeper understanding of the clinical implications of load carriage and its association with spinal stability and spinal motor control, we propose to investigate the changes of motor control of spine during posterior load carriage. The findings are valuable to the field of scientific and clinical research to explore the effects of posterior load carriage, and it may help to establish the guidelines for load carriage usage. Besides these, this study also aims to apply high-heeled shoes to counteract the effects of load carriage.

We sincerely invite you to participate in this study. A detailed information sheet is attached with this letter for you together with more information about this study. Please sign the consent form when you agree to participate in the study. The study will be conducted in the Ergonomics Laboratory of The Hong Kong Polytechnic University. Your spine curvature and dynamical response during load carriage will be documented using non-invasive measurements (external device attached to your skin surface). The load carriage and the device would not induce any discomfort to you. The experiment will be terminated immediately if you report any discomfort and the incidence will be recorded.

Information sheet and consent form are attached with this letter. For further information or queries, please contact Prof. Daniel Chow (Tel.: 27667674) or Mr. WANG Chao (Tel.: 27664361).

Prof. Daniel Chow

Interdisciplinary Division of Biomedical Engineering (BME)

The Hong Kong Polytechnic University

Part b:

Information sheet

Research Title: Effects of Load Carriage and High-Heeled Shoes on Spinal Motor Control

We sincerely invite you to participate in this study conducted by the Department of Health Technology and Informatics of The Hong Kong Polytechnic University. The research objective and experimental procedures are described in detail in this information sheet. Please read it carefully before joining this study and please feel free to contact us for further information or enquiry.

Investigator

Mr. WANG Chao

Interdisciplinary Division of Biomedical Engineering (BME)

The Hong Kong Polytechnic University

Supervisor

Professor Daniel HK Chow

Interdisciplinary Division of Biomedical Engineering (BME)

The Hong Kong Polytechnic University

Purpose of Research

Backpack is an everyday use article for many people. Many studies have shown that heavy load carriage would cause muscle soreness, and may lead to low back pain. However, the effects of load carriage on spinal motor control are still not clear. The aim of this study is to determine the changes of spinal motor control due to posterior load carriage. The results will be useful for reducing the risk of spine injury or low back pain due to load carriage. The other objective of this study is to explore the possibility of using high-heeled shoes to counteract the effect of load carriage

Procedures

The experiment will be carried out in the Hong Kong Polytechnic University. All participants are requested to sign a consent form to show that they have understood the purpose and procedures of the research.

1. Participant's name, age, body weight and height will be recorded.
2. During the experiment, the participant has to wear a pair of shorts. The male subjects required exposing their upper body and female subjects are required to wear corsage only.
3. Several sensors will be attached to the skin surface of the participant's back for data collection.
4. Data will be collected under several weights of load carriage (i.e. 0, 5%, 10% and 15%BW) combined with different heel heights (i.e. barefoot, flat shoes, 2cm high heel shoes and 5cm high heel shoes). For each condition, the participant will be asked to perform a functional reaching task. The participant will be instructed to reach forward using his trunk and hips as if he is reaching over a counter into a cupboard, touch a stationary target and immediately return to the upright standing posture. This task will be repeated three times consecutively.

The whole experiment will last for about two hours.

Risks and Discomforts

There is no risk involved in this study. Mild muscle soreness may be developed after the test.

Confidentiality

You will only be asked for general information about sex, age, height, weight, and the history of past medical health. The result obtained in this study may be published while all the personal information will be kept confidential.

Your participation in this study is entirely voluntary and you may refuse to participate or withdraw from the study at any time without any explanation.

Contact for Information about the Study

If you have any questions with respect to this study, please contact Professor Daniel Chow (Tel: 27667674) or Mr. WANG Chao (Tel: 27664361). If you have any complaints about the conduct of this research study, please contact Ms Kath Lui, Secretary of the Human Subject Ethics Sub-Committee of The Hong Kong Polytechnic University in person or in writing (c/o Research Office of the University). Thank you very much for reading this information sheet and considering to participate in the study.

Part c:

Consent Form

Research Title: Effects of Load Carriage and High-Heeled Shoes on Spinal Motor Control

Investigator

Mr. WANG Chao

Interdisciplinary Division of Biomedical Engineering (BME)

The Hong Kong Polytechnic University

Supervisor

Professor Daniel HK Chow

Interdisciplinary Division of Biomedical Engineering (BME)

The Hong Kong Polytechnic University

（please ✓ as appropriate）

1. I have read and understand the contents on the information sheet, and I have the right to inquire about the study. ☐
2. My participation in this study is entirely voluntary and I may refuse to participate or withdraw from the study at any time without any punishment. ☐
3. I understand the results of this study may be reviewed by other researches, I agree them to access my record. ☐
4. I agree to participate in above study. ☐

Participant's name: ____

- Signature of participant: ____________________
- Name of participant: _______________________
- Date: _______________

- Signature of witness（if necessary）: ____________________
- Name of witness: _______________________
- Date: _______________

- Signature of investigator: ____________________
- Name of witness: _______________________
- Date: _______________

Chinese Version:

邀請您參與《背包與高跟鞋對脊柱運動控制功能的影響》研究

腰背痛是非常普遍的，大約 80%的人都曾經歷過腰背痛。有證據顯示，復發性的腰背痛與脊柱運動控制功能受損有關。腰背痛的病人有軀幹肌肉反應延遲的現象，而這個反應延遲被證明會增加腰背受傷和背痛復發的機會。背包亦被發現為增加學童患腰背痛的機會的一個危險因素，這亦會增加他們日後患上慢性腰背痛的可能。以前的研究結果表明背包會明顯影響脊柱彎曲度和損害脊柱的定位能力。因此，為了更深入地探究背包對脊柱穩定性和脊柱運動控制功能影響的臨床意義，我們致力於研究背上背包時，脊柱運動控制功能的改變。該研究的結果將會對在科學和臨床上繼續探究背包的影響有重要作用，而且可能有助於建立背包使用的指導性意見。除此之外,本研究還將嘗試用高跟鞋去抵消書包對脊柱帶來的影響。

我們誠邀您參與此項研究。請仔細閱讀參考資料。如果你同意參與，請簽署同意書。此研究將會在香港理工大學人體工效學實驗室進行。背上背包後的脊柱弧度、動態反應將會用無創的方法（貼在皮膚表面的感應器）測量。負重和測量設備不會對您造成任何不適。一旦您感到任何不適，該實驗會立刻終止，而且該狀況亦會被記錄。

隨信附上參考資料和同意書，如有任何疑問和查詢，請聯絡周教授(電話.: 2766 7674 or 電子郵箱: Daniel.Chow@) 或王超先生 （電話：27664361，電子郵箱：htchao.wang@ ）

周鴻奇教授

香港理工大學生物醫學工程跨領域學部

研究項目資料

背包與高跟鞋對脊柱運動控制功能的影響

我們誠邀您參與一項由香港理工大學醫療科技及資訊學系進行的研究。這份參考資料詳細介紹了該項研究的研究目的、實驗步驟以及其他相關的事宜。在您決定是否參加這項研究之前，請仔細閱讀此份資料。如果有任何問題或諮詢，請隨時與我們聯絡。

研究員：王超先生（香港理工大學生物醫學工程跨領域學部）

導師：周鴻奇教授（香港理工大學生物醫學工程跨領域學部）

研究目的：

背包是一件普及的日常用品。雖然很多研究指出，過重的背包會使肌肉產生疲勞，並有可能誘發腰背痛等不良反應。但背包對脊柱本身運動控制功能的影響尚未確定。這項研究的目的正在於提供關於攜帶背包對脊柱運動控制能力影響之數據，從而避免使用背包時產生脊柱損傷和可能引致的腰背痛。本研究的另外一個目的是探索用高跟鞋抵消書包對脊柱影響的可能性。

程序：

該項研究將安排在香港理工大學進行，此次研究的所有參加者均需簽署研究同意書表明已明白此項研究的目的和方法。

研究須知：

1. 研究人員將記錄參加者的姓名、年齡、身高和體重。
2. 進行研究時，參加者下身需穿著短褲，男士上身不穿衣服/女士穿緊身泳衣。
3. 參加者腰背部皮膚表層需被貼上若干感應器以進行脊柱數據收集（固有展露上半身之需要）。
4. 脊柱數據將會在參加者於不同負重條件下收集。測試包括：無負重及背負重量為 5%,10%和 15%體重的背包,與赤腳,平底鞋,2cm 高跟鞋,5cm 高跟鞋的不同組合。於這些負重狀況下參加者將分別做以下測試：在研究員的指導下，連續完成一個類似彎腰伸手拾物的動作三次。

整個測試約持續兩個小時。

潛在風險與不適

整個測試過程危險性極低；在測試後，您可能會有輕微肌肉酸痛現象。

資料保密

您只需要提供如性別、年齡、身高、體重和健康狀況的資料。在發佈該研究的研究成果時，所有有關您的個人資料將會被嚴格保密。

參與是自願的，您可以無需提供任何理由的情況下，隨時退出此項研究。

查詢

如對此項研究有任何疑問，可與周鴻奇教授(電話:27667674) 或王超先生 (電話: 27664361) 聯絡。如對施行本研究的過程有任何投訴，請以親身聯絡或致函至：香港理工大學研究事務處，道德標準小組委員會秘書呂小姐。多謝閣下閱讀此份資料和考慮參與此項研究。

<u>同意書</u>

研究題目：背包與高跟鞋對脊柱運動控制功能的影響

研究員姓名：王超先生（香港理工大學生物醫學工程跨領域學部）

導師姓名：周鴻奇教授（香港理工大學生物醫學工程跨領域學部）

請填上"✓"號

1. 我已閱讀並理解參考資料的有關內容,
   並且了解我有提出問題的權利。 ☐

2. 我明白我的參與是自願的，我可以在任何時間自由退出而不必給予任何理 由，
   而且我不遭受任何處罰。 ☐

3. 我明白有關我參與這項研究的有關記錄可能被其他相關的研究員檢閱,
   我批准該等人士獲取我的記錄。 ☐

4. 我同意參加以上的研究。 ☐

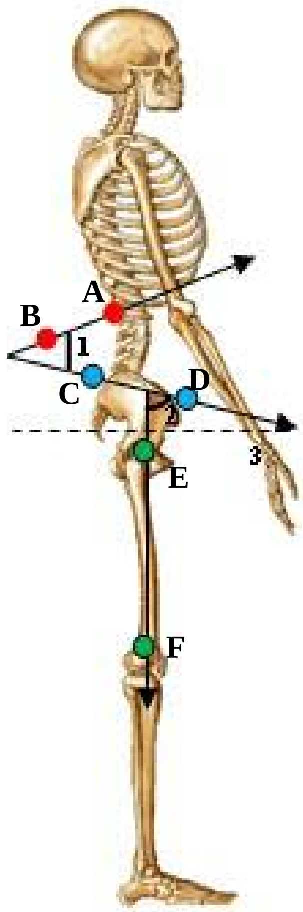
參加者姓名和身份證號碼: ____________________

- 參加者簽署：____________________
- 參加者姓名: ____________________
- 日期：____________________

- 見證人簽署（如需要）：____________________
- 見證人姓名: ____________________
- 日期：____________________

- 研究員簽署：____________________
- 研究員姓名: ____________________
- 日期：____________________

Appendix 2: Maker Placement and Calculation



<u>Markers</u>

Middle point of both sides of L1: Point A

Protruded L1: Point B

Middle Point of PSIS: Point C

Middle Point of ASIS: Point D

Right Great Trochanter: Point E

Right Knee: Point F

Vectors

Vector BA= $\vec{L}$

Vector CD= $\vec{H}$

Vector EF= $\vec{T}$

$$\vec{L} = \begin{bmatrix} L_x \\ L_y \\ L_z \end{bmatrix} = \begin{bmatrix} A_x - B_x \\ A_y - B_y \\ A_z - B_z \end{bmatrix} \qquad \vec{H} = \begin{bmatrix} H_x \\ H_y \\ H_z \end{bmatrix} = \begin{bmatrix} D_x - C_x \\ D_y - C_y \\ D_z - C_z \end{bmatrix} \qquad \vec{T} = \begin{bmatrix} T_x \\ T_y \\ T_z \end{bmatrix} = \begin{bmatrix} F_x - E_x \\ F_y - E_y \\ F_z - E_z \end{bmatrix}$$

Equations

$$\vec{L} \bullet \vec{H} = |\vec{L}||\vec{H}|\cos\theta_1$$

$$\cos\theta_1 = \frac{(A_x - B_x)(D_x - C_x) + (A_z - B_z)(D_z - C_z)}{\sqrt{(A_x - B_x)^2 + (A_z - B_z)^2}\ \sqrt{(D_x - C_x)^2 + (D_z - C_z)^2}}$$

$$\theta_1 = \cos^{-1}\frac{(A_x - B_x)(D_x - C_x) + (A_z - B_z)(D_z - C_z)}{\sqrt{(A_x - B_x)^2 + (A_z - B_z)^2}\ \sqrt{(D_x - C_x)^2 + (D_z - C_z)^2}}$$

$$\vec{H} \bullet \vec{T} = |\vec{H}||\vec{T}|\cos\theta_2$$

$$\cos\theta_2 = \frac{(D_x - C_x)(F_x - E_x) + (D_z - C_z)(F_z - E_z)}{\sqrt{(D_x - C_x)^2 + (D_z - C_z)^2}\ \sqrt{(F_x - E_x)^2 + (F_z - E_z)^2}}$$

$$\theta_2 = \cos^{-1}\frac{(D_x - C_x)(F_x - E_x) + (D_z - C_z)(F_z - E_z)}{\sqrt{(D_x - C_x)^2 + (D_z - C_z)^2}\ \sqrt{(F_x - E_x)^2 + (F_z - E_z)^2}}$$

$$\tan\theta_3 = \frac{C_z - D_z}{D_x - C_x}$$

$$\theta_3 = \tan^{-1}\frac{C_z - D_z}{D_x - C_x}$$

Appendix 3: Statistical Analysis Results

A3.1 Statistical result for main effect and their interactions for study of effect of load carriage only (Phase I)

A3.1.1: Summary of statistical results about the main effects and their interactions for lumbar initial upright posture

	Lumbar Initial Upright Posture
Loading	p=0.319
Gender	p=0.140
Loading * Gender	p=0.372

(*: significant effect)

A3.1.2: Summary of statistical results about the main effects and their interactions for pelvis initial upright posture

	Pelvis Initial Upright Posture
Loading	p=0.574
Gender	p=0.096
Loading * Gender	p=0.240

(*: significant effect)

A3.1.3: Summary of statistical results about the main effects and their interactions for lumbar repositioning error

	Lumbar Repositioning Error
Loading	p=0.651
Gender	p=0.148
Loading * Gender	p=0.589

(*: significant effect)

A3.1.4: Summary of statistical results about the main effects and their interactions for pelvis repositioning error

	Pelvis Repositioning Error
Loading	p=0.865
Gender	p=0.062
Loading * Gender	p=0.880

(*: significant effect)

A3.1.5: Summary of statistical results about the main effects and their interactions for reaching distance

	Reaching Distance
Loading	p<0.001 *
Gender	p=0.067
Loading * Gender	p=0.126

(*: significant effect)

A3.1.6: Summary of statistical results about the main effects and their interactions for mean absolute relative phase (MARP)

	MARP
Loading	p<0.001 *
Movement Direction	p=0.001 *
Gender	p=0.028 *
Loading * Gender	p=0.245
Loading * Movement Direction	p=0.354
Movement Direction * Gender	p=0.495
Loading * Movement Direction * Gender	p=0.411

(*: significant effect)

A3.1.7: Summary of statistical results about the main effects and their interactions for deviation phase (DP)

	DP
Loading	p<0.001 *
Movement Direction	p=0.372
Gender	p=0.193
Loading * Gender	p=0.827
Loading * Movement Direction	p=0.031 *
Movement Direction * Gender	p=0.607
Loading * Movement Direction * Gender	p=0.474

(*: significant effect)

A3.1.8: Summary of statistical results about the main effects and their interactions for movement ratio

	Movement ratio
Loading	p=0.001 *
Gender	p=0.981
Loading * Gender	p=0.386

(*: significant effect)

A3.2 Pairwise comparison among various carrying load conditions for study of effect of load carriage only (Phase I)

A3.2.1: p-value of pairwise comparison for lumbar initial upright posture

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.338			
10%BW	0.578	0.085		
15%BW	0.498	0.128	0.814	

A3.2.2: p-value of pairwise comparison for pelvis initial upright posture

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.245			
10%BW	0.859	0.291		
15%BW	0.206	0.799	0.377	

A3.2.3: p-value of pairwise comparison for lumbar repositioning error

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.626			
10%BW	0.511	0.356		
15%BW	0.596	0.222	0.913	

A3.2.4: p-value of pairwise comparison for pelvis repositioning error

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.849			
10%BW	0.809	0.722		
15%BW	0.440	0.442	0.686	

A3.2.5: p-value of pairwise comparison for reaching distance

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	**0.001**			
10%BW	**<0.001**	**<0.001**		
15%BW	**<0.001**	**<0.001**	**0.005**	

A3.2.6: p-value of pairwise comparison for mean absolute relative phase (MARP)

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.736			
10%BW	**0.004**	**0.007**		
15%BW	**<0.001**	**0.001**	**0.044**	

A3.2.7: p-value of pairwise comparison for deviation phase (DP) during forward motion

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.436			
10%BW	**0.002**	**0.013**		
15%BW	**<0.001**	**0.001**	**0.010**	

A3.2.8: p-value of pairwise comparison for deviation phase (DP) during backward motion

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.883			
10%BW	0.165	0.103		
15%BW	**<0.001**	**0.001**	**0.004**	

A3.2.9: p-value of pairwise comparison for lumbar movement ratio

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.254			
10%BW	**0.008**	0.051		
15%BW	**0.005**	**0.026**	0.192	

A3.3 Statistical result for main effect and their interactions for study of combined effects of load carriage and high-heeled shoes (Phase II)

A3.3.1: Summary of statistical results about the main effects and their interactions for lumbar initial upright posture

	Lumbar Initial Upright Posture
Loading Heel Height Loading * Heel Height	p=0.518 p=0.012 * p=0.588

(*: significant effect)

A3.3.2: Summary of statistical results about the main effects and their interactions for pelvis initial upright posture

	Pelvis Initial Upright Posture
Loading Heel Height Loading * Heel Height	p=0.077 p=0.004 * p=0.754

(*: significant effect)

A3.3.3: Summary of statistical results about the main effects and their interactions for lumbar repositioning error

	Lumbar Repositioning Error
Loading Heel Height Loading * Heel Height	p=0.048 * p=0.058 p=0.490

(*: significant effect)

A3.3.4: Summary of statistical results about the main effects and their interactions for pelvis repositioning error

	Pelvis Repositioning Error
Loading Heel Height Loading * Heel Height	p=0.143 p=0.050 * p=0.424

(*: significant effect)

A3.3.5: Summary of statistical results about the main effects and their interactions for reaching distance

	Reaching Distance
Loading	p<0.001 *
Heel Height	p<0.001 *
Loading * Heel Height	p=0.740

(*: significant effect)

A3.3.6: Summary of statistical results about the main effects and their interactions for mean absolute relative phase (MARP)

	MARP
Loading	p=0.006 *
Movement Direction	p=0.002 *
Heel Height	p<0.001 *
Loading * Heel Height	p=0.638
Loading * Movement Direction	p=0.290
Movement Direction * Heel Height	p=0.223
Loading * Movement Direction * Heel Height	p=0.958

(*: significant effect)

A3.3.7: Summary of statistical results about the main effects and their interactions for deviation phase (DP)

	DP
Loading	p=0.030 *
Movement Direction	p=0.600
Heel Height	p=0.001 *
Loading * Heel Height	p=0.863
Loading * Movement Direction	p=0.251
Movement Direction * Heel Height	p=0.435
Loading * Movement Direction * Heel Height	p=0.481

(*: significant effect)

A3.3.8: Summary of statistical results about the main effects and their interactions for movement ratio

	Movement ratio
Loading	p=0.636
Heel Height	p<0.001 *
Loading * Heel Height	p=0.449

(*: significant effect)

A3.4 Pairwise comparison among various carrying load and heel height conditions for study of combined effects of load carriage and high-heeled shoes (Phase II)

A3.4.1: p-value of pairwise comparison for lumbar initial upright posture

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.085			
10%BW	0.711	0.419		
15%BW	0.198	0.796	0.477	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	0.131			
5cm High-heeled shoes	**0.009**	0.073		

A3.4.2: p-value of pairwise comparison for pelvis initial upright posture

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.074			
10%BW	0.089	0.243		
15%BW	0.062	0.197	0.941	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	**0.010**			
5cm High-heeled shoes	**0.004**	0.272		

A3.4.3: p-value of pairwise comparison for lumbar repositioning error

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	**0.033**			
10%BW	**0.031**	0.240		
15%BW	**0.035**	0.373	0.808	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	0.058			
5cm High-heeled shoes	0.051	0.620		

A3.4.4: p-value of pairwise comparison for pelvis repositioning error

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.053			
10%BW	0.072	0.814		
15%BW	0.118	0.648	0.763	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	0.063			
5cm High-heeled shoes	**0.050**	0.785		

A3.4.5: p-value of pairwise comparison for reaching distance

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	**<0.001**			
10%BW	**<0.001**	0.073		
15%BW	**<0.001**	**0.022**	0.475	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	**<0.001**			
5cm High-heeled shoes	**<0.001**	**0.002**		

A3.4.6: p-value of pairwise comparison for mean absolute relative phase (MARP)

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.151			
10%BW	0.064	0.607		
15%BW	**0.011**	0.487	0.828	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	**0.003**			
5cm High-heeled shoes	**0.000**	**0.021**		

A3.4.7: p-value of pairwise comparison for deviation phase (DP)

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.221			
10%BW	0.051	0.186		
15%BW	**0.004**	0.133	0.978	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	**0.001**			
5cm High-heeled shoes	**0.003**	0.202		

A3.4.8: p-value of pairwise comparison for lumbar movement ratio

	NOBP	5%BW	10%BW	15%BW
NOBP				
5%BW	0.366			
10%BW	0.273	0.828		
15%BW	0.369	0.990	0.874	
	Flat shoes	2cm High-heeled shoes	5cm High-heeled shoes	
Flat shoes				
2cm High-heeled shoes	**<0.001**			
5cm High-heeled shoes	**<0.001**	**0.022**		

Appendix 4 Raw Data of Each Parameter

Table A4.1 The Participants' information

Subject number	Gender	Age (year)	Body height (cm)	Body weight (kg)
1	Male	26	174	68
2	Male	28	176	72.7
3	Male	23	172	69.2
4	Male	30	170	58.4
5	Male	28	179	68.8
6	Male	29	173	80.8
7	Male	24	171	65.9
8	Male	28	172	58.8
9	Female	22	166	79.6
10	Female	26	166	60.7
11	Female	26	150	49
12	Female	27	167	52.3
13	Female	23	160	42.5
14	Female	28	160	60
15	Female	26	159	56.5
16	Female	25	154	49.7
17	Female	21	156	45.4
18	Female	21	157	53.1
19	Female	20	150	54.7
20	Female	21	158	52.5
21	Female	19	158	42.2
22	Female	21	167	58.3
23	Female	19	158	58
24	Female	21	161	46.6
25	Female	24	162	46.2
26	Female	22	160	46.4
27	Female	21	164	54.2
28	Female	21	150	52.7

A4.2 Raw data of the study of effects of load carriage only

Table A4.1.1 Raw data of the functional reaching distance

No.	Gender	Functional reaching distance (cm)			
		NOBP	5%BW	10%BW	15%BW
1	Male	38.0	34.0	32.0	31.7
2	Male	40.7	37.3	34.3	30.7
3	Male	44.0	37.7	34.0	30.3
4	Male	41.7	37.7	33.7	27.7
5	Male	46.0	41.0	40.3	35.7
6	Male	45.7	43.3	40.0	35.0
7	Male	43.7	40.0	37.7	37.3
8	Male	39.0	36.0	36.3	30.3
9	Female	35.0	36.7	30.0	27.7
10	Female	40.7	40.3	34.0	31.0
11	Female	42.0	41.3	36.0	35.7
12	Female	39.0	31.3	32.0	29.7
13	Female	40.0	38.0	32.0	28.0
14	Female	37.0	37.3	32.7	37.0
15	Female	37.3	36.0	31.0	28.0
16	Female	39.3	35.0	32.0	31.0

Table A4.2.2 Raw data of initial upright lumbar posture / lumbar repositioning error

No.	Gender	Initial upright lumbar posture / Lumbar repositioning error (°)			
		NOBP	5%BW	10%BW	15%BW
1	Male	15.3/1.4	16.6/1.7	17.9/1.4	18.4/2.9
2	Male	31.7/1.2	32.8/1.0	30.2/1.4	32.2/0.9
3	Male	34.4/1.0	33.4/2.0	35.2/1.1	35.2/1.1
4	Male	20.1/1.9	22.5/0.9	24.9/0.6	23.3/0.5
5	Male	38.9/1.3	39.3/2.8	37.7/1.9	40.4/1.8
6	Male	15.6/0.9	18.5/1.1	17.1/1.4	17.4/1.1
7	Male	16.5/1.4	17.9/1.3	18.1/1.2	13.2/1.8
8	Male	31.8/1.1	16.0/0.8	16.8/0.8	28.1/0.8
9	Female	18.2/1.1	19.2/0.5	21.0/1.0	20.7/1.7
10	Female	20.8/0.7	18.5/1.9	35.9/1.3	31.6/1.2
11	Female	15.3/1.9	18.1/1.9	29.3/1.7	21.3/1.9
12	Female	18.5/4.1	20.2/3.0	21.2/4.1	16.8/2.3
13	Female	21.8/2.1	20.3/2.7	19.0/1.1	23.6/1.9
14	Female	20.2/1.6	15.1/0.8	16.5/1.7	15.5/0.5
15	Female	26.1/1.2	20.3/2.5	21.0/1.8	22.6/1.6
16	Female	12.5/1.8	9.9/1.4	12.1/1.8	9.4/1.1

Table A4.2.3 Raw data of initial upright pelvic tilting / pelvic repositioning error

No.	Gender	Initial upright pelvic tilting / Pelvic repositioning error (°)			
		NOBP	5%BW	10%BW	15%BW
1	Male	4.5/2.3	2.9/1.8	3.8/1.4	3.3/2.9
2	Male	-1.7/1.2	-2.1/1.2	0.2/1.4	-0.5/1.2
3	Male	-7.7/0.6	-6.2/1.5	-7.9/0.8	-4.9/1.5
4	Male	8.9/1.1	7.7/0.6	6.0/0.8	8.8/0.6
5	Male	11.0/0.8	-10.6/0.9	-6.1/1.1	-8.6/0.8
6	Male	9.8/1.5	9.8/1.5	11.7/0.8	10.7/1.5
7	Male	9.0/1.12	8.2/0.8	6.2/1.1	9.1/0.6
8	Male	-1.8/0.9	12.0/0.8	11.1/1.3	0.0/0.9
9	Female	4.4/0.9	4.7/0.8	3.9/1.3	4.1/0.9
10	Female	4.2/1.1	7.7/1.6	-9.2/1.7	-2.3/1.5
11	Female	19.7/1.9	16.6/2.0	4.4/1.6	19.1/2.0
12	Female	14.2/4.2	13.8/1.9	14.2/4.2	17.6/3.1
13	Female	5.8/2.8	6.8/3.2	8.2/1.2	6.4/1.6
14	Female	7.5/1.7	8.8/1.1	9.0/1.6	10.4/1.4
15	Female	-0.1/1.0	1.7/2.5	4.2/1.5	1.7/1.5
16	Female	12.2/1.6	14.7/1.7	13.5/1.4	17.0/1.4

Table A4.2.4 Raw data of the forward mean absolute relative phase

No.	**Gender**	**Forward Mean Absolute Relative Phase (MARP_Forward) (°)**			
		NOBP	**5%BW**	**10%BW**	**15%BW**
1	Male	18.3	27.5	23.8	24.9
2	Male	18.1	20.0	22.8	40.2
3	Male	20.0	18.7	23.7	58.3
4	Male	24.5	24.5	30.5	30.2
5	Male	9.7	13.0	27.7	18.5
6	Male	23.1	18.6	26.1	30.0
7	Male	26.8	24.1	41.5	45.1
8	Male	18.6	23.8	34.1	56.2
9	Female	20.8	24.5	33.1	46.3
10	Female	19.7	11.2	16.9	21.9
11	Female	7.2	15.6	19.7	24.2
12	Female	22.9	37.6	28.0	20.9
13	Female	18.2	15.2	15.8	33.9
14	Female	27.0	17.7	30.2	31.3
15	Female	13.1	10.6	14.7	9.8
16	Female	8.5	7.6	9.7	21.4

Table A4.2.5 Raw data of backward mean absolute relative phase

No.	Gender	Backward Mean Absolute Relative Phase (MARP_Backward) (°)			
		NOBP	5%BW	10%BW	15%BW
1	Male	34.5	38.7	36.4	40.4
2	Male	28.9	30.6	37.5	50.9
3	Male	16.6	16.4	17.0	47.4
4	Male	28.3	29.4	39.5	45.8
5	Male	14.6	18.4	42.6	24.3
6	Male	31.6	24.1	29.6	32.8
7	Male	36.7	31.9	51.1	43.1
8	Male	30.3	25.5	34.6	46.7
9	Female	26.8	28.7	41.4	48.0
10	Female	23.4	17.3	20.0	27.7
11	Female	13.7	21.1	25.7	20.5
12	Female	31.8	41.3	33.0	30.5
13	Female	22.5	19.7	16.8	34.4
14	Female	22.2	20.3	28.2	28.2
15	Female	22.0	22.9	18.4	22.4
16	Female	12.1	10.4	12.4	28.7

Table A4.2.6 Raw data of the forward deviation phase

No.	Gender	Forward Deviation Phase (DP_Forward) (°)			
		NOBP	5%BW	10%BW	15%BW
1	Male	15.6	22.4	20.2	23.9
2	Male	17.4	19.9	21.8	31.6
3	Male	15.8	20.0	21.4	38.4
4	Male	14.4	16.2	19.1	23.0
5	Male	10.1	11.3	22.3	17.5
6	Male	15.9	12.7	19.6	20.8
7	Male	22.2	18.8	26.2	29.3
8	Male	17.9	16.6	18.4	36.4
9	Female	16.5	17.4	27.2	35.4
10	Female	17.6	12.4	14.9	19.9
11	Female	6.8	12.9	16.3	21.2
12	Female	19.2	34.0	27.7	19.5
13	Female	13.0	10.7	12.0	23.5
14	Female	17.4	14.6	24.3	26.9
15	Female	13.4	11.6	11.2	10.5
16	Female	9.1	7.5	10.6	19.4

Table A4.2.7 Raw data of backward deviation phase

No.	Gender	Backward Deviation Phase (DP_Forward) (°)			
		NOBP	**5%BW**	**10%BW**	**15%BW**
1	Male	24.7	27.7	26.0	21.9
2	Male	19.2	24.0	26.2	32.4
3	Male	13.9	14.8	14.9	25.7
4	Male	10.7	13.1	15.1	19.7
5	Male	12.1	12.6	12.2	19.7
6	Male	16.0	13.9	21.4	18.4
7	Male	22.8	14.4	19.7	27.8
8	Male	18.5	12.8	13.2	24.2
9	Female	13.7	13.8	21.6	29.8
10	Female	19.0	16.3	16.0	20.6
11	Female	12.9	13.4	17.5	15.3
12	Female	18.2	28.2	26.4	24.0
13	Female	8.2	8.3	6.7	17.4
14	Female	11.0	14.2	18.7	19.4
15	Female	20.9	20.5	15.2	20.5
16	Female	12.0	8.5	11.8	21.8

Table A4.2.8 Raw data of lumbar movement ratio

No.	Gender	Lumbar movement ratio			
		NOBP	5%BW	10%BW	15%BW
1	Male	0.57	0.63	0.60	0.59
2	Male	0.55	0.63	0.62	0.63
3	Male	0.59	0.52	0.55	0.70
4	Male	0.57	0.56	0.57	0.57
5	Male	0.49	0.54	0.50	0.48
6	Male	0.61	0.60	0.65	0.63
7	Male	0.64	0.61	0.69	0.67
8	Male	0.55	0.66	0.67	0.74
9	Female	0.57	0.64	0.66	0.68
10	Female	0.57	0.53	0.59	0.57
11	Female	0.51	0.54	0.59	0.64
12	Female	0.50	0.56	0.52	0.55
13	Female	0.64	0.62	0.63	0.70
14	Female	0.66	0.58	0.62	0.61
15	Female	0.49	0.51	0.55	0.51
16	Female	0.48	0.53	0.55	0.58

A4.3 Raw data of the study of combined effects of load carriage and high-heeled shoes

Table A4.3.1 Raw data of the functional reaching distance

No.	Gender	Functional reaching distance (cm)											
		BP01 HH01	**BP01 HH02**	**BP01 HH03**	**BP02 HH01**	**BP02 HH02**	**BP02 HH03**	**BP03 HH01**	**BP03 HH02**	**BP03 HH03**	**BP04 HH01**	**BP04 HH02**	**BP04 HH03**
1	F	43.3	35.3	30.7	37.3	32.0	29.3	34.0	28.0	28.3	30.3	30.0	30.3
2	F	34.0	31.3	28.7	29.0	30.3	28.7	31.0	25.7	26.7	23.3	24.3	19.0
3	F	32.3	29.7	29.3	28.7	26.7	26.3	28.0	25.3	27.3	33.3	26.7	19.3
4	F	37.3	35.3	30.7	33.7	30.0	26.7	32.7	32.0	29.0	28.3	25.7	26.7
5	F	30.7	26.7	22.3	22.0	20.0	23.3	20.3	21.0	24.3	24.7	23.0	26.0
6	F	33.3	27.7	29.3	32.7	29.3	25.7	25.0	20.3	22.7	30.3	27.3	24.7
7	F	38.0	36.0	33.3	34.3	34.0	30.7	35.0	31.3	31.0	33.0	31.0	32.0
8	F	42.0	37.0	33.7	36.0	28.3	28.7	28.0	26.0	19.7	31.3	22.7	24.3
9	F	25.0	22.3	23.0	26.3	24.3	23.0	25.7	21.7	20.7	23.7	25.0	20.3
10	F	39.7	34.0	36.0	36.3	29.3	30.0	31.0	35.0	28.7	27.0	28.0	23.7
11	F	35.3	33.3	35.0	31.7	30.3	28.0	30.0	28.7	27.7	30.3	28.0	25.3
12	F	35.0	34.7	39.0	29.3	32.0	27.7	29.3	33.0	33.0	35.3	29.7	29.3

Table A4.3.2 Raw data of initial upright lumbar posture / lumbar repositioning error (°)

No.	Gender	Initial upright lumbar posture / Lumbar repositioning error (°)											
		BP01 HH01	**BP01 HH02**	**BP01 HH03**	**BP02 HH01**	**BP02 HH02**	**BP02 HH03**	**BP03 HH01**	**BP03 HH02**	**BP03 HH03**	**BP04 HH01**	**BP04 HH02**	**BP04 HH03**
1	F	21.4/ 2.5	28.4/ 1.6	28.7/ 2.0	22.2/ 1.1	24.1/ 1.9	24.0/ 2.3	25.2/ 1.9	33.4/ 2.1	29.3/ 2.2	30.8/ 2.0	32.3/ 3.2	27.7/ 3.8
2	F	19.2/ 2.4	15.8/ 2.1	17.6/ 4.4	13.9/ 3.2	10.8/ 3.2	14.6/ 4.8	14.3/ 2.2	13.2/ 3.3	12.1/ 3.3	10.3/ 2.1	16.9/ 4.7	16.2/ 3.4
3	F	15.8/ 0.8	16.2/ 2.51	20.3/ 2.3	15.4/ 0.8	18.0/ 2.4	18.7/ 2.3	18.1/ 1.6	16.2/ 2.5	20.4/ 1.4	17.6/ 1.7	20.2/ 0.7	22.2/ 0.7
4	F	27.1/ 1.6	22.0/ 3.7	27.1/ 1.3	22.9/ 2.3	16.2/ 1.9	22.8/ 2.6	25.0/ 4.0	21.1/ 3.0	21.0/ 4.5	26.4/ 3.8	22.8/ 4.3	26.0/ 3.2
5	F	22.4/ 0.4	27.2/ 0.6	28.6/ 0.7	25.8/ 0.4	26.4/ 0.8	28.6/ 1.0	25.1/ 1.1	26.9/ 0.6	29.4/ 0.4	24.6/ 0.3	26.1/ 0.8	26.7/ 0.8
6	F	13.1/ 1.5	18.8/ 1.5	24.2/ 2.4	16.5/ 1.3	18.6/ 2.6	23.3/ 1.5	15.3/ 1.6	18.3/ 3.2	20.2/ 1.4	17.7/ 2.2	18.6/ 1.1	17.4/ 5.0
7	F	11.1/ 1.3	19.6/ 1.0	21.4/ 1.5	17.0/ 1.0	15.6/ 1.3	21.5/ 1.3	11.7/ 1.2	12.7/ 1.0	14.3/ 2.1	12.9/ 1.0	11.0/ 0.7	14.9/ 1.8
8	F	22.8/ 1.1	23.7/ 1.4	25.3/ 0.8	21.9/ 2.5	25.1/ 0.5	28.4/ 2.2	27.9/ 1.2	30.2/ 5.3	28.8/ 1.3	23.5/ 1.7	26.5/ 1.0	24.2/ 1.9
9	F	47.0/ 0.5	46.3/ 0.3	46.1/ 0.4	48.9/ 0.7	45.5/ 0.4	44.4/ 0.2	46.8/ 0.8	46.8/ 0.4	47.1/ 0.4	49.9/ 0.4	45.3/ 0.3	44.0/ 0.3
10	F	14.0/ 0.8	15.6/ 0.8	19.4/ 0.7	12.6/ 0.5	14.5/ 1.4	9.8/ 1.7	10.0/ 1.3	9.4/ 0.6	12.7/ 1.4	10.9/ 1.2	14.6/ 2.2	13.4/ 1.3
11	F	20.5/ 2.0	22.6/ 2.4	21.3/ 1.2	20.2/ 2.0	18.8/ 2.6	20.9/ 2.9	16.1/ 2.4	19.8/ 5.8	19.4/ 2.7	17.6/ 3.3	17.9/ 1.1	21.5/ 2.3
12	F	36.7/ 0.9	36.3/ 0.3	35.7/ 0.7	35.9/ 1.8	38.6/ 0.9	33.5/ 1.4	32.1/ 1.4	46.8/ 0.4	47.1/ 0.4	10.8/ 1.2	31.9/ 1.0	32.3/ 1.8

Table A4.3.3 Raw data of initial upright pelvic tilting / pelvic repositioning error (°)

No.	Gender	Initial upright pelvic tilting / Pelvic repositioning error (°)											
		BP01 HH01	**BP01 HH02**	**BP01 HH03**	**BP02 HH01**	**BP02 HH02**	**BP02 HH03**	**BP03 HH01**	**BP03 HH02**	**BP03 HH03**	**BP04 HH01**	**BP04 HH02**	**BP04 HH03**
1	F	6.2/ 2.4	3.4/ 1.4	2.7/ 1.4	3.9/ 1.0	2.9/ 1.2	2.3/ 1.7	2.0/ 1.2	-0.9/ 2.1	1.8/ 1.2	2.1/ 1.5	1.7/ 2.5	3.6/ 3.2
2	F	10.2/ 1.3	11.7/ 2.2	12.0/ 4.1	17.2/ 3.4	17.6/ 2.8	13.6/ 5.1	15.4/ 2.5	15.8/ 3.3	16.6/ 2.8	19.2/ 2.2	14.5/ 5.1	17.1/ 3.2
3	F	10.9/ 1.6	10.4/ 2.7	7.8/ 2.5	12.0/ 1.3	8.7/ 2.7	9.5/ 2.5	11.2/ 2.5	9.0/ 2.7	8.2/ 1.4	8.4/ 2.1	10.5/ 0.8	9.2/ 0.9
4	F	16.2/ 1.9	19.1/ 3.3	14.7/ 1.5	19.0/ 2.7	16.6/ 5.4	18.8/ 2.6	15.6/ 3.7	18.5/ 3.0	18.6/ 4.5	17.4/ 3.8	16.3/ 6.0	10.7/ 7.3
5	F	0.8/ 0.5	-1.6/ 0.7	-1.5/ 1.0	-0.5/ 0.6	-0.9/ 0.8	-1.8/ 1.1	-0.9/ 1.3	-2.5/ 0.4	-2.1/ 0.7	-0.7/ 0.5	0.4/ 1.3	-1.6/ 0.6
6	F	7.0/ 1.9	3.0/ 1.2	0.9/ 2.0	4.1/ 1.2	4.5/ 1.9	2.1/ 1.9	6.4/ 1.7	6.1/ 3.4	4.4/ 1.2	6.5/ 2.0	4.6/ 1.3	5.8/ 4.5
7	F	11.4/ 1.1	6.7/ 1.2	6.0/ 1.4	7.6/ 2.1	8.4/ 1.3	6.7/ 1.4	10.9/ 0.7	12.2/ 0.7	9.8/ 2.0	10.8/ 1.0	12.4/ 0.8	10.4/ 1.7
8	F	10.5/ 1.1	9.8/ 1.7	8.4/ 0.8	12.4/ 2.3	11.2/ 0.8	9.2/ 2.4	10.9/ 1.1	7.7/ 6.0	11.4/ 1.1	11.1/ 1.2	10.1/ 1.0	11.2/ 1.4
9	F	9.1/ 0.7	9.0/ 0.4	8.9/ 0.4	7.5/ 0.8	9.0/ 0.6	9.6/ 0.4	9.0/ 1.0	7.8/ 0.6	7.6/ 0.4	5.7/ 0.4	9.1/ 0.4	10.2/ 0.4
10	F	18.2/ 1.1	16.5/ 0.8	15.2/ 1.0	20.5/ 0.8	19.3/ 1.2	15.3/ 1.1	22.8/ 1.6	20.8/ 0.7	19.5/ 1.9	20.3/ 1.2	19.2/ 1.8	20.2/ 1.5
11	F	10.5/ 1.3	10.5/ 1.9	10.4/ 1.4	12.2/ 1.7	9.6/ 4.0	12.3/ 3.3	13.4/ 2.4	12.1/ 5.5	12.7/ 2.2	12.7/ 3.2	14.6/ 1.4	12.6/ 2.2
12	F	-3.1/ 1.2	-3.7/ 0.8	-3.3/ 1.0	-0.6/ 1.6	-3.7/ 0.9	3.1/ 1.3	5.3/ 1.4	7.8/ 0.6	7.6/ 0.4	10.3/ 1.2	3.6/ 0.7	3.4/ 1.7

Table A4.3.4 Raw data of forward mean absolute relative phase

No.	Gender	Forward Mean Absolute Relative Phase (MARP_Forward) (°)											
		BP01 HH01	**BP01 HH02**	**BP01 HH03**	**BP02 HH01**	**BP02 HH02**	**BP02 HH03**	**BP03 HH01**	**BP03 HH02**	**BP03 HH03**	**BP04 HH01**	**BP04 HH02**	**BP04 HH03**
1	F	28.6	18.8	7.9	8.5	4.6	4.6	8.4	7.0	8.7	15.5	15.9	15.2
2	F	8.5	8.4	13.8	23.7	12.0	13.0	30.6	21.2	11.5	13.7	24.9	14.7
3	F	26.8	20.9	22.9	19.7	23.6	27.9	27.5	19.8	27.1	27.1	30.7	37.7
4	F	11.5	6.2	6.0	17.9	9.9	7.3	12.1	16.2	8.9	9.4	21.4	15.4
5	F	29.9	24.8	29.3	29.5	26.8	22.2	33.6	28.1	29.0	42.8	34.2	42.3
6	F	30.5	37.1	34.1	38.2	40.0	40.7	44.2	34.5	31.0	43.9	27.8	30.8
7	F	11.7	9.1	13.4	23.2	12.9	15.9	14.8	19.8	11.4	24.8	21.3	12.3
8	F	15.6	18.9	17.0	18.7	17.0	15.8	37.7	30.9	26.3	24.9	16.9	10.7
9	F	16.4	14.4	27.6	54.8	43.2	21.2	30.3	28.3	32.4	57.4	22.8	21.2
10	F	29.6	25.5	29.3	41.9	33.2	26.8	39.0	25.8	21.3	30.4	38.8	30.1
11	F	14.4	10.5	9.5	12.1	12.9	14.3	13.1	12.9	9.3	14.9	11.4	14.2
12	F	27.2	16.0	19.3	43.4	32.4	18.4	49.4	28.4	32.4	30.4	27.4	23.9

Table A4.3.5 Raw data of backward mean absolute relative phase

No.	Gender	Backward Mean Absolute Relative Phase (MARP_Backward) (°)											
		BP01 HH01	BP01 HH02	BP01 HH03	BP02 HH01	BP02 HH02	BP02 HH03	BP03 HH01	BP03 HH02	BP03 HH03	BP04 HH01	BP04 HH02	BP04 HH03
1	F	36.4	25.6	15.1	16.4	8.3	6.4	12.5	11.7	11.9	23.3	21.8	19.0
2	F	8.1	12.6	17.9	23.1	12.8	14.4	29.4	25.1	11.6	12.8	26.7	18.9
3	F	43.9	30.1	33.8	38.6	31.2	39.4	46.3	31.9	40.8	41.0	45.9	49.8
4	F	21.1	8.5	10.6	26.6	15.6	15.8	20.6	24.2	9.6	17.2	25.8	21.7
5	F	34.6	36.8	40.8	42.3	36.2	24.9	45.6	34.1	26.1	49.0	45.2	32.9
6	F	50.5	49.1	46.1	45.4	62.6	49.6	45.1	47.4	51.1	45.8	43.6	35.9
7	F	16.9	14.9	13.3	20.7	16.6	16.4	15.7	17.0	12.0	25.7	18.1	19.5
8	F	26.1	30.9	29.7	35.0	33.3	21.8	52.1	33.7	31.5	34.6	24.2	15.8
9	F	16.3	15.5	26.4	55.1	44.7	24.3	38.4	32.5	31.9	56.2	25.4	20.8
10	F	37.0	27.8	27.8	40.2	38.3	33.1	37.5	37.4	23.2	37.1	34.9	19.9
11	F	25.6	24.0	12.9	26.4	23.6	32.5	25.8	31.3	24.3	25.3	25.0	32.1
12	F	29.3	18.3	19.9	33.8	25.7	21.2	45.0	32.5	31.9	37.1	23.2	25.2

Table A4.3.6 Raw data of forward deviation phase

No.	Gender	Forward Deviation Phase (DP_Forward) (°)											
		BP01 HH01	BP01 HH02	BP01 HH03	BP02 HH01	BP02 HH02	BP02 HH03	BP03 HH01	BP03 HH02	BP03 HH03	BP04 HH01	BP04 HH02	BP04 HH03
1	F	25.6	22.1	8.4	7.7	3.9	4.6	8.6	6.5	9.3	13.7	15.5	16.1
2	F	7.6	8.9	14.6	21.7	12.9	11.8	25.1	16.7	12.7	12.4	21.0	15.7
3	F	24.3	19.1	21.8	15.8	23.1	25.2	24.6	18.3	25.7	27.0	27.0	32.9
4	F	11.9	6.4	6.1	19.0	10.0	7.0	11.1	15.2	8.6	9.5	20.0	14.0
5	F	20.8	17.5	20.0	21.1	17.9	15.2	23.1	18.8	23.2	25.9	25.2	32.1
6	F	20.4	25.5	24.7	25.1	29.0	29.1	30.1	26.4	23.8	32.4	21.6	21.9
7	F	10.2	8.6	13.8	23.4	11.9	16.7	13.0	16.6	9.7	21.2	22.7	11.6
8	F	14.2	16.4	12.5	16.1	15.8	15.7	32.0	24.9	25.5	20.5	17.1	10.4
9	F	14.7	10.3	25.2	32.0	33.2	20.5	23.1	26.4	27.6	36.2	20.9	19.0
10	F	20.0	18.8	20.1	28.8	22.6	18.8	30.5	22.2	18.0	22.8	26.6	21.4
11	F	13.4	10.6	9.9	12.2	14.5	15.3	12.2	15.6	10.0	16.1	12.2	14.7
12	F	19.4	13.8	15.0	25.4	22.1	17.1	29.7	26.4	27.6	22.8	19.6	19.6

Table A4.3.7 Raw data of backward deviation phase

No.	Gender	Backward Deviation Phase (DP_Backward) (°)											
		BP01 HH01	BP01 HH02	BP01 HH03	BP02 HH01	BP02 HH02	BP02 HH03	BP03 HH01	BP03 HH02	BP03 HH03	BP04 HH01	BP04 HH02	BP04 HH03
1	F	21.9	18.5	12.7	12.2	5.6	4.4	8.9	9.1	9.5	12.7	18.4	14.5
2	F	7.3	9.7	14.9	18.2	11.5	12.6	19.8	13.6	9.5	10.2	21.0	17.4
3	F	30.2	23.3	22.4	29.1	23.2	28.8	29.5	24.1	25.9	20.6	29.6	35.7
4	F	19.3	9.3	11.0	14.3	13.7	17.2	16.1	17.9	9.5	16.0	17.8	17.0
5	F	19.0	19.9	19.0	17.9	13.5	15.5	16.9	18.1	16.0	18.9	21.9	20.4
6	F	32.5	32.1	29.1	27.7	26.3	36.2	33.8	30.9	29.8	26.1	23.3	27.8
7	F	12.2	13.5	12.6	18.1	15.9	16.3	12.1	13.5	10.0	17.8	17.2	20.1
8	F	22.1	23.7	19.5	24.8	22.2	19.3	35.3	29.2	27.4	27.8	20.7	14.7
9	F	14.2	10.1	19.7	32.7	29.9	20.0	25.0	26.5	28.3	35.3	22.0	17.4
10	F	15.0	13.9	15.5	18.9	19.7	21.3	26.2	22.8	20.6	22.4	22.1	15.2
11	F	13.7	13.4	9.2	19.7	14.4	15.5	13.0	23.0	15.8	16.6	20.4	15.4
12	F	16.0	10.7	11.1	16.5	14.3	12.3	20.0	26.5	28.2	22.4	15.0	13.4

Table A4.3.8 Raw data of lumbar movement ratio

No.	Gender	Lumbar movement ratio											
		BP01 HH01	BP01 HH02	BP01 HH03	BP02 HH01	BP02 HH02	BP02 HH03	BP03 HH01	BP03 HH02	BP03 HH03	BP04 HH01	BP04 HH02	BP04 HH03
1	F	0.63	0.56	0.50	0.53	0.52	0.50	0.52	0.48	0.50	0.53	0.54	0.53
2	F	0.52	0.48	0.46	0.60	0.50	0.51	0.56	0.52	0.50	0.53	0.55	0.52
3	F	0.67	0.60	0.57	0.65	0.57	0.62	0.65	0.59	0.53	0.55	0.62	0.66
4	F	0.55	0.54	0.55	0.53	0.53	0.52	0.56	0.56	0.53	0.55	0.62	0.56
5	F	0.57	0.52	0.53	0.61	0.53	0.53	0.59	0.56	0.52	0.70	0.63	0.54
6	F	0.59	0.61	0.69	0.63	0.55	0.57	0.65	0.56	0.55	0.57	0.50	0.55
7	F	0.46	0.43	0.41	0.52	0.48	0.43	0.48	0.55	0.45	0.54	0.48	0.48
8	F	0.59	0.60	0.57	0.58	0.55	0.56	0.60	0.63	0.60	0.59	0.54	0.53
9	F	0.65	0.61	0.62	0.80	0.78	0.66	0.64	0.69	0.64	0.83	0.62	0.57
10	F	0.51	0.50	0.52	0.56	0.51	0.53	0.61	0.54	0.52	0.55	0.60	0.53
11	F	0.52	0.49	0.49	0.51	0.49	0.49	0.50	0.49	0.47	0.52	0.50	0.48
12	F	0.59	0.52	0.57	0.70	0.60	0.54	0.68	0.69	0.64	0.55	0.56	0.55

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