A biomechanical approach to investigate the effects on the lumbosacral joint, pelvis, and knee joint while of carrying asymmetrical loads, during ground walking.

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ABSTRACT

Background: Spinal pain is reasonably considered among the expensive and impairing problems critically disturbing the health of individuals, especially the workforce, in industrially developed countries (Steele et al., 2003). The pain adversely affecting the lumbar region or pelvis is typically considered as low back pain. As per the National Institute of Neurological Disorder and Stroke, around 80 percent of grown-ups encounter spinal pain eventually in the course of their lifetime. The possible risk factors include age, fitness level, genetics, weight gain, occupational risk factors such as having a job that expects someone to do lifting, carrying, pushing, pulling, distorting the spinal column, and backpack overload in children, etc.(Low Back Pain Fact Sheet | National Institute of Neurological Disorders and Stroke, n.d.). Many lifting studies and backpack carrying studies have adequately identified carrying a load as an activity that contributes to possible risk to the problems in the lumbar spine (Cholewicki et al., 1991; Daniel H.K. Chow et al., 2005; Goh et al., 1998). Most of the published studies have investigated school-going children, recreational hikers, or military personnel with a heavy backpack. However, the present literature lacks the information of precisely locating the lumbosacral joint coordinate system and the possible effects on the lumbosacral joint while carrying a one strap electrical and maintenance tool bag which moves or swings side to side as the carrier moves. Backpacks, used for carrying books, stationery items, or items used by recreational hikers, are typically stable but the one strap tool bag has to be suspended from the shoulder. This suspension may result in the swing of the bag from side to side, or it may strike the lateral side of the pelvis while walking. Even if the carrier wants to stabilize the tool bag, he/she will have to hold the bag while walking, which results in restricting the arm swing. Another way to carry the tool bag is to hold the handle of the tool bag if it has any. Even in this case, the arm will have limited swing. Therefore, this study aims to develop a method

that can investigate the effects on the spine, pelvis, and knee from carrying loads asymmetrically (on one shoulder and in one hand) during ground walking.

Objective: The objective of this experiment is to develop a method for the evaluation of kinetics and kinematics variables to assess the strain level of the participant while carrying asymmetrical loads during ground walking. This method aims to

- Identify and analyze the Lumbosacral joint (L5-S1) disc compression and shear force
- Identify and analyze the pelvis obliquity, tilt, and rotation
- Identify and analyze the knee joint compression

CONTENTS

CHAPTER ONE: INTRODUCTION

Most of the research that have been done on load carrying is based on school-going children who carry heavy backpacks, or recreational hikers who carry heavy loads during backpacking. Commonly, backpacks are considered to be the fundamental medium of load-carrying equipment and this form of load carriage varies based on the need for carrying(Ismaila, 2018). Students from different age groups use backpacks for carrying books, notebooks, other stationeries, and even laptops, while hikers use backpacks for carrying their hiking supplies like tents, fast aids, torches, water bottles, etc. Variation in heaviness and duration of usage depends on the ways backpacks are used (Al-Khabbaz et al., 2008). Carrying a heavy load in a backpack has been one of the primary causes of short term and/or long-term musculoskeletal disorders for persons from different age groups. A close relationship between musculoskeletal disorders and heavy backpack carriage exists in growing school children (Shamsoddini et al., 2010). Improper and heavy carriage of a backpack can lead to long term musculoskeletal disorders on neck, shoulder, and back in school going children (Alsiddiky et al., 2019). Even walking for a short period with a backpack can significantly change the spinal curvature(Orloff & Rapp, 2004). Increased loading on spinal tissue and negative effects on adolescent spinal responses have been found due to inappropriate and prolonged load carriage(Grimmer & Williams, 2000). Each year, around 13000 severe injuries are reported that are associated with backpacks in the US, according to the U.S. Consumer Product Safety Commission (2015). A significant amount of research has been done by researchers all over the world to find out the safe weight limit to be carried. Almost all the researchers advocated a safe weight limit between 10%-15% bodyweight depending on epidemiological, physiological, and biomechanical approaches (Brackley & Stevenson, 2004). A study was conducted on young adults considering the number of straps of a backpack as a factor to find out if this factor could be a contributor to musculoskeletal discomfort or alteration of the gait parameters. The results showed no influence on the parameters of the gait cycle, but only an increase in perceived exertion for both strapping conditions (Abaraogu et al., 2016). Improper distribution in weights causes abnormal postures from a biomechanical perspective (Gong et al., 2010).There has been a substantial amount of research done on school-going children, military personnel, and hikers who walk a significant amount of time carrying their respective backpacks, but there are almost no studies found on carrying electrical tool bag unilaterally. The following section gives a review of the studies on carrying backpacks and other load carriage systems.

2 CHAPTER TWO: LITERATURE REVIEW

2.1 GAIT

Several studies have been performed to investigate the effects of load carriage on gait parameters. These include the effects due to different load carriage methods, load conditions, strap patterns, and load positions. The results were, based mostly on children and military personnel, were not all in agreement.

Load Condition: Carrying a backpack directly affects some of the gait parameters of a person compared to walking without a backpack (Cottalorda et al., 2003; Singh & Koh, 2009). In several studies (Cottalorda et al., 2003; Majumdar et al., 2010; Qu & Yeo, 2011) significant increases in step width, stride length, step length, cadence, midstance, stance, and double stance were observed when walking with a heavy backpack compared with walking without a backpack. Other studies (Harman et al., 2000; Martin & Nelson, 1986; Winter et al., 1990) found significantly different gait patterns under different load conditions; They found a decrease in both the stride length and the swing time with increasing load. They also observed increased stride rate and double support time as the load carried by the participant increased. Kinoshita (1985) also found a significant increase in double support time as load increased in the backpack (Kinoshita, 1985). The author suggested that step length should be shortened with the increase in load for the faster transfer of the body weight from one leg to another. Kinoshita also found that the body posture and gait patterns were nearer to normal walking while using the double-pack system (front/back-packing system).

Some studies, however, found no significant differences in gait parameters while walking with a backpack with increasing loads compared to walking without a backpack(Abaraogu et al., 2016; Connolly et al., 2008; Hong & Cheung, 2003; Krupenevich et al., 2015).

Strap Pattern: In a recent study, Abaraogu et al. 2016 investigated the effect of backpack strap pattern on gait parameters during the self-determined fast walking of young adults. They found a significant decrease in stride time and cadence during young adults walk fast with a load of up to 10% of body weight while wearing a double strap backpack. However, there were no significant effects on the gait parameters at normal walking speed when the load was less than 20% of body weight (Abaraogu et al., 2016).

Load Position: Several studies investigated the impact of load carriage and its posterior position on gait parameters and observed that a placing the load low on the back results in a reduced gate velocity and an increased double support time compared to placing loads higher. (Daniel H.K. Chow et al., 2005; Singh & Koh, 2009)

Pal et al. (2009) conducted a study on Indian infantry soldiers to determine an optimal load to be carried at two different walking velocities. The study results recommend 36.1 kg and 21.3 kg as permissible load carriage at a speed of 3.5 and 4.5km/hr respectively on level ground for comfortable walking (Pal et al., 2009).

Therefore, based on the previous studies discussed above, several inconsistencies among the results of the available studies are apparent and most of the studies were conducted on children, recreational hikers or trained military personnel. However, most of the studies agree with the changes in spatiotemporal parameters of gait while walking with load carriage systems.

2.2 GROUND REACTION FORCE

Ground reaction force (GRF) is generated by the ground on an object or body that stays in contact with the ground. In this case, this reaction force is distributed across the whole area of the contact between the feet and the floor. The ground reaction force is an important external force, along with the weight. GRF is considerably associated with anterior shear force acting on the proximal tibia, making this a risk factor for Anterior Cruciate Ligament injury, though lesser in males than females. In this experiment, the suspension of the tool bag over one shoulder made the bag sway from side to side and made walking uncomfortable. The swing of the bag disturbed the general balance of the human gait. Analysis of the effects of restriction or elimination of arm swing on the components of the peak ground reaction force should, therefore, yield important information on the risk to injury from this kind of bag carrying. The ground reaction force allowed calculation of forces and moments at each joint of the body.

Ground reaction force (GRF) at the feet-surface contact is expected to increase, vertically, forward, rearward, and laterally, as load increases (J. Knapik et al., 1996). Most, but not all, studies have confirmed this experimentally (S. A. Birrell & Haslam, 2008; Majumdar et al., 2013; Miletello et al., 2008; Peduzzi de Castro et al., 2014) but differences may be due to factors other than mere load weight during the carrying. Birrell et al. (2008) found an increase in the vertical GRF during the stance phase while carrying a rifle compared with no rifle (S. A. Birrell & Haslam, 2008); Miletello et al. (2008) in a study of ten college students found that a backpack load of 20% body weight, in a stair descent test, increased vertical GRF by 29.5% during the stance phase and increased it by 15.38% during toe-off phase, compared to no load (Miletello et al., 2008) ; and Majumdar et al. (2013) found an increase in vertical GRF in a study among Indian infantry soldiers (Majumdar et al., 2013). However, Birell et al. (2010) in a study with military personnel using three different load carriage systems (backpack, standard and AirMesh), found a reduction in the GRF at toe-off and a significant decrease in the stance time for heavier loads compared to lighter loads. The authors suggested that this was due to the shifting of center of mass of the loads posteriorly (Stewart A. Birrell & Haslam, 2010). Peduzzi de Castro et al.(2014) found an increase in absolute GRF with load in backpack carriage, but the normalized force (relative to total weight) showed a reduction in GRF with load weight (Peduzzi de Castro et al., 2014). Majumder et al. (2013), however, found that normalization did not change the relation between GRF and load weight, in the study on Indian infantry soldiers (Majumdar et al., 2013). Forces needed for maintaining balance increase significantly when the load is carried. However, Peduzzi de Castro et al. (2014) found that the influence of gait speed on kinetic gait parameters, such as shear and vertical forces and plantar pressures, during load carriage is not consistent. Tillbury-Davis &

Hooper, (1999) found that the increase in GRF was proportional to the pack weight, but there was no significant increase in knee flexion with load carriage, and also no significant difference in gait parameters. (Tilbury-Davis & Hooper, 1999).

2.3 Lumbosacral Joint Force

The lumbosacral joint, also called the L5-S1 spinal motion segment, is the transition area between the lumbar spine and sacral spine in the lower back. This joint helps transfer loads from the spine into the pelvis and legs. This joint receives a higher degree of mechanical load than the segments above. Because of these characteristics, the L5-S1 DISC is considered susceptible to traumatic injuries, including degeneration, herniation, or nerve pain. As mentioned in the literature review there is not a significant amount of literature on the L5-S1 disc compression and shear forces from walking and carrying a load on the shoulder with a restricted arm swing. Though this joint has a high strength to support both the axial and shear forces generated from load-carrying activities, it is not known how these forces change over from asymmetric carrying on one shoulder and in one hand. This study tried to yield the relevant data to make these assessments.

Walking with load posteriorly induces added mechanical stress on the spine and it poses a very significant amount threat to the lower back. A study by Li et al., 2019 investigated the effect of walking with backpack loads (5%, 10%, 20%, and 25% of Bodyweight) on the lumbosacral joint compression force. (Li et al., 2019). Ten male adults participated in this study. A disproportionate escalation of force profiles was observed as the backpack load increased. The effects of increasing the backpack loads became more noticeable as the peak lumbosacral joint force increased due to the increase in load carried by the subjects. This type of increase in joint compression force could be due to the change in posture of different body segments and alterations in the activation of muscles. It was noteworthy that the percentage increase in the average L5-S1 joint compression forces were not proportional to the percentage increase in the backpack loads.

McGill et al., 2013 investigated the effects of carrying a load unilaterally versus bilaterally. (McGill et al., 2013). Unilateral load carrying generated more compressive force at the lower back than that of the bilateral mode of load carrying. The low back compression was more than 2800 N during the unilateral load carriage. However, carrying the same amount of load bilaterally resulted in a 44% reduction in the low back compression force. The compression force on the low back was lower even when the participants carried 60kg bilaterally compared to carrying 30kg unilaterally.

A study by Wang et al., 2017 investigated the effects on the lumbosacral joint while carrying unilateral and bilateral weight conditions in the instance of stair negotiation (J. Wang & Gillette, 2017). Part of the study's objective was to explore the effects of both the above-mentioned weight conditions on the lumbosacral joint. The study recruited healthy young adults as participants and the participants performed the experimental tasks with five different load conditions. Load condition included no load, a symmetrical load of 10% body weight, an asymmetrical load of 10% body weight, a symmetrical load of 20% body weight, and an asymmetrical load of 20% body weight. The lateral bending moment at L5-S1 disc was significantly higher while carrying an asymmetric load of 20% bodyweight compared to other load conditions. Moreover, lateral bending moments while carrying an asymmetrical load of 10% body weight were higher compared to no load condition.

Tilbury-Davis & Hooper, 1999 conducted a study on 10 military subjects that resulted in proportional increase in ground reaction forces as the load carried by the participant increased and the peak forces needed to facilitate the stabilization of the gait due to the additional weight carried by the participant increased significantly (Tilbury-Davis & Hooper, 1999). Lloyd & Cooke, 2000 assessed the changes in kinetics from a no-load walking condition with the loaded condition using a traditional backpack and an innovative load carriage system with front balance pockets (Lloyd & Cooke, 2000). They did not find the increase in anteroposterior forces to be proportional to the system weight as reported by other studies. In addition to this finding, the increase in the propulsive force while carrying the new load carriage system (front/back system) was significantly smaller than the traditional backpack, as load increased. This observation recommends that, having a front/back system as a load carriage medium could be more advantageous than the traditional back system when it comes to propulsive force generation. Increases in the vertical ground reaction forces, as load increased, were found to be proportional to the loads carried by the participants. Overall, this study (Lloyd & Cooke, 2000) intends to support the likely advantages of the front/back system based on the generation of reduced required anteroposterior propulsive force.

2.4 PHYSIOLOGICAL PARAMETERS

Datta and Ramanathan, 1971 investigated different modes carrying, including head, rucksack, double pack, rice nag, Sherpa, yoke, and hand and found that the double pack system was ergonomically the best method based on some physiological factors such as energy cost, cardiac rate, and pulmonary ventilation (Datta & Ramanathan, 1971). Walking comprises a form of repeatedly losing and regaining balance, since during walking the center of gravity moves continually outside the base of support and more importantly the feet are never flat on the ground at the same time. In a study by Soule et al., 1978, when subjects walked with loads from 35-70 kg at a different speed, the net energy expenditure was constant at each speed and showed no statistically significant difference (Soule et al., 1978). This constancy depends on the condition that the load is well balanced and closer to the center of the body. Repetitive load carriage for a longer period is associated with musculoskeletal disorders including knee pain, foot blisters, stress fractures and back strain. J.J. Knapik et al, 2004 reviewed the historical and biomedical aspects of soldier load carriage (J. J. Knapik et al., 2004). From literature it is evident that positioning the center of mass of the load carriage system close to the center of mass of the body could lower the energy cost and this likewise will in general keep the body-load system in an upstanding position.

Legg et al., 1992, in a study, with subjects carrying a twenty-six kg load (two -part) on each shoulder as one load carriage condition and attached to a framed backpack as another condition, found that the relative oxygen cost was lower for the later condition than for shoulder condition. (Legg et al., 1992). The authors concluded that the metabolic cost associated with asymmetrical load carriage in a framed backpack was significantly lower than that of shoulder carriage during walking. In an experiment, carrying load wearing lightweight athletic shoes and heavier boots, the energy expenditure was seen to be increased by wearing boots. The large contribution to that increased energy expenditure is caused by the boot weight. The increase in energy cost is assumed to be putting a stress on the subjects who are not trained (Jones et al., 1984). A study by Winter, 1983 showed contradictions with the previous claims of increased energy cost with rigid knee (Winter, 1983). A significant correlation between energy cost and knee flexion was evident as an increase in energy cost was reported with the increase in knee flexion angle. A study conducted by Legg and Mahanty (1985) aimed to investigate the physiological reactions during carrying load keeping the load carriage system close to the trunk (Legg & Mahanty, 1985).

Five subjects participated in that study. Each carried a load of 35% of their body weight for an hour and walked at a predetermined speed on a treadmill. The load carriage conditions included: (I) load carried inside a framed backpack (2) load carried inside a no frame backpack (3) 50% of the load in a framed backpack and 50% attached to a waist belt (4) 50% of load inside a framed backpack and 50% inside a front pack and (5) as a trunk jacket. Oxygen uptake (V02), minute ventilation (V) and heart rate (HR) were measured every 10min during each experimental run. At 35% of body weight condition, the study reported no significant changes in cardiorespiratory and metabolic cost in all the different modes of load carrying. J Knapik & Reynolds, 1996 reviewed the biomedical aspects of transporting loads and suggested some points that might improve the load carriage capability (J. Knapik & Reynolds, 1997). As mention earlier, the CoM of the load should be placed as close as possible to the CoM of the body as this might help in lowering the energy cost. Thus, the normal backpack has a higher energy cost than the double pack (equal distribution of the load carried by the frontal and back part of the body). Compared to backpacks, double packs produce fewer deviations from normal walking. B.-S., 2007 studied the effect of backpack load position, walking speed and surface grade on the physiological responses of infantry soldiers and found walking speed as strongly significant for all physiological indices (B.-S., 2007). Load carried at the upper position of the pack resulted in significantly higher mean respiratory frequency and mean oxygen consumption. However, carrying heavy loads close to the trunk can affect lung function. Therefore, it is suggested to take the lung function into consideration as well in future studies.

2.5 JOINT KINETICS AND KINEMATICS (KNEE, HIP, PELVIS AND ANKLE)

Load carriage carriage had different effects on the lower extremities on different stages of a gait cycle. Inconsistencies among the kinematics variables such as joint angles of hip, ankle, and knee have been seen on the studies conducted to the date. Most of the biomechanical experiment associated with load carriage have been performed during ground walking and most of the cases the investigators considered symmetrical load carriage. However, very limited research on asymmetrical load carriage has been conducted to report the biomechanical alteration of joint kinetics and kinematics of hip, knee, and ankle. The knowledge of kinetics and kinematic alterations in hip, knee, and ankle requirements due to external loads need to be considered for future applications such as in the design of lower limb prostheses, orthopedic and neurological rehabilitation. This section considers both the symmetric and asymmetric load carriage as the present study also tried to gather the knowledge on how the increased load carriage affects the joint kinetics and kinematics of lower extremities.

2.5.1 Knee Flexion/Extension Moment

Knee osteoarthritis is associated with increased external adduction moment during walking (Hall et al., 2017) so it is of value to understand how load-carrying methods impact these moments. Previous findings suggest increased medial knee joint moment while carrying two-strap backpacks with increasing load. One study found a significant relation between knee moment and backpack type (two-strap backpack and backTpack) and load (Dahl et al., 2016). Remarkably, while carrying 15% of body weight load, knee moments were greater for backTpack than the backpack, but at 25% body weight condition the knee moments for the backpack got greater than backTpack. Further investigations on the knee moments affected by increased load are needed since developing knee osteoarthritis might have a correlation with increased knee moments. Knee force analysis is very significant in understanding osteoarthritis, and the survival and function of knee arthroplasty. Quesada et al., 2000 in a study on 12 military personnel walking with backpack loads of 0%, 15% and 30% of body weight observed that the knee flexion moment increased by about 82% and 151% during stance phase at 15% and 30% of BW load, respectively (Quesada et al., 2000). Other studies (Brown et al., 2014; Lee et al., 2017; Majumdar et al., 2010) had similar results of increased knee flexion moment at heel strike as the backpack load increased. However, a significant overall increase in the knee extension moment with increasing backpack load was observed in some studies (D. H.K. Chow et al., 2007; Krupenevich et al., 2015; H. Wang et al., 2013) as well. This variation of significant knee flexion/extension moment during initial contact could be a result of the differences in populations sampled in those studies. Wang et. al., 2013 and Chow et al., 2007 recruited adult students and schoolgirls respectively, whereas the other studies, mentioned earlier, recruited a mixture of young adults and military personnel. Therefore, load carriage during walking has a noteworthy effect on the knee joint moment. This increased knee joint moment might be responsible for providing increased total weight support and shock absorption.

2.5.2 Ankle dorsi/plantar flexion moment

Researchers observed an increase in ankle dorsiflexion and plantarflexion during different phases of a gait(Lee et al., 2017). A significant increase in the dorsiflexion moment at the ankle during the stance phase was reported with the increased load carriage in several studies (Quesada et al., 2000). The force generated from the posterior weight could be responsible for pushing the ankle into greater dorsiflexion. Increase ankle plantarflexion moment was also reported in some studies (Krupenevich et al., 2015; Quesada et al., 2000) during the early to midstance phase with the increase of load carriage. The ankle plantar flexors play a major role in human locomotion.

The increase might have a relation with the elastic energy in the tendons of ankle during walking (Farris & Sawicki, 2012). Therefore, the force of the tendon during walking with heavy load posteriorly could be greater than during walking with none or lighter loads. However, one study reported unchanged ankle plantarflexion moment (H. Wang et al., 2013).

2.5.3 Hip Flexion/Extension Moment

A close review on the existing literature indicates that the load carriage during walking is associated with altered hip moment. The center of the gravity above the hip shifts backward with increased load in the backpack which forces the net torque at hips to alter. This might happen as a compensatory mechanism to stabilize the load at the joints of the hip. This mechanism might also reduce the necessity of bending the body caused by increased load at the back. During the period between late stance and toe-off, an increase in hip extension internal moment (Quesada et al., 2000; H. Wang et al., 2013) and increased hip flexor moment (Krupenevich et al., 2015) was observed and this might be due to the additional backpack load. On the contrary, several studies documented no change at the hip moments during walking as the load of backpack increased (Huang & Kuo, 2014; Krupenevich et al., 2015; Majumdar et al., 2010). The magnitude of forward trunk flexion angle may be a determining factor in this discrepancy of varied hip moments. The subjects of Krupenvich et al. had significant trunk flexion angle during walking when a backpack was carried whereas other studies (Quesada et al., 2000; H. Wang et al., 2013) did not observe any trunk flexion angle.

2.5.4 Knee Flexion/Extension angle

There are studies that observed variation in knee flexion/extension angles at different points of a gait during walking with load carriage. Between early to midstance, several studies have observed increased (Kinoshita, 1985; Simpson et al., 2012; H. Wang et al., 2013) and unchanged (Ghori & Luckwill, 1985; Majumdar et al., 2010) peak knee flexion angle during walking with an increase in backpack load. There are other evidence through studies (P. et al., 2004; Skaggs et al., 2006) that support the theory that the load carried posteriorly can contribute to increased knee flexion angle during level walking. This increased flexion angle may have a relation with the shock absorption and compensation mechanism for extra weight. During terminal stance to pre-swing, both reduced peak knee flexion (Kinoshita, 1985)but no change in peak knee flexion angle were reported in different studies (Majumdar et al., 2010; Simpson et al., 2012). The reason could be that Kinoshita tested healthy male subjects who were not regularly engaged in carrying tasks, whereas the participants of Majumder et al., 2010 and Simpson et al., 2012 were military personnel and professional recreational hikers respectively.

2.5.5 Ankle Dorsi/Plantar Flexion angle

Studies observing the effect of increased load carriage on ankle dorsiflexion and plantar flexion showed inconsistencies among their result. Some studies (Majumdar et al., 2013; H. Wang et al., 2013) found no significant changes in the ankle dorsiflexion angle with the increased weight of backpack carriage. However, there was evidence of increase in both ankle plantarflexion (Push-off) and dorsiflexion angles (Stance phase), as load carriage increased, at different points of the gait cycle (Clark et al., 2018). This increase in ankle dorsiflexion might be due to load carriage system held posteriorly. However, the increase in ankle plantar flexion might be considered as a compensatory mechanism due to the additional weight applied to the back and this increased range of motion might require to move forward (Clark et al., 2018; Kinoshita, 1985; Quesada et al., 2000; Simpson et al., 2012). One study (Kinoshita, 1985) observed reduced ankle dorsiflexion angle as load carriage increased. This difference might be due to the sample of participants Kinoshita observed as they were all healthy male people with no prior experience in carrying heavy loads.

2.5.6 Pelvis Angle

The literature indicates that carrying a heavy bag might contribute to the permanent changes in posture which might result in high compressive force in the lumbosacral joint and compensatory changes in pelvic motion (tilt, obliquity and rotation) that alters gait. Few studies that have analyzed the biomechanical compensation of the pelvis while carrying a one strap tool bag suspended over one shoulder with restricted arm swing. Different studies could not concur on joint hip angles during different phases of the gait cycle. During initial contact to loading response phase of a gait cycle, an increased (H. Wang et al., 2013) and unchanged (Majumdar et al., 2013) hip flexion angle was found with load carriage during ground walking. No change in hip extension was observed by (Majumdar et al., 2013) between early to midstance phase. Backpack carriage was associated with an increase in hip extension angle (Kinoshita, 1985; Majumdar et al., 2013) during terminal to pre-swing phase. However, similar studies with increasing load carriage in backpack, reported reduced (Caron et al., 2013) and unchanged (Quesada et al., 2000) hip extension angle. The increase in hip flexion angle is due to increased moments and power at the hip joint to compensate for the increasing demand with backpack load. The decrease in hip extension angle during the terminal to pre-swing phase might occur as a part of the contribution to stabilize the alerted pelvic due to the increased load in the back. Based on the findings in the above literature review, the present study was conducted to develop a more accurate method to assess the biomechanics of asymmetric arm carrying (on one side of the body, on the shoulder or in the hand).

3 CHAPTER 3: METHODOLOGY

The purpose of this study was to develop a method to investigate the kinetic and kinematic effects on the spine, pelvis, and knee from carrying loads asymmetrically (on one shoulder or in the hand) while walking. Carrying heavy loads regularly over the years is associated with frequent back pain and other musculoskeletal disorders. In this experiment, an electrical tool bag of a specific brand was used as the load carriage system and the participants had to walk carrying the bag with different loads according to their body weight. As stated in the literature review in Chapter 1, there are many investigations regarding load carriage systems, mostly with backpacks, and their effects on joint kinetics and kinematics of different segments of the human body. The load carriage system used in this study was different from regular backpacks. It was an electrical and maintenance tool bag with only one strap and one handle for carrying. Therefore, it could be carried either by suspending it over the shoulder by the strap, or by holding the handle in one hand. In this experiment, we tried to study both of these modes of carrying. A laboratory pilot experiment in carrying bag over the shoulder showed that the tool bag swayed from side to side of the participant. The one-sided lateral location of the bag results in additional side-to-side movement while walking, however, the typical backpacks moves from front to back. To stop this lateral movement, the participants used one hand to stabilize the bag while walking, resulting in a restricted arm swing. This seems to be normal among people. A pilot experiment was conducted to evaluate the duration and feasibility of the experiments, the possible adverse events, subject preparation, and data collection instruments associated with the study design. The swing restriction of one arm and the tendency of the bag to sway from side-to-side are likely to change the effects on the kinetic and kinematic variables from bag carrying that. This study will assess these changes.

The variables investigated were carrying mode (two levels) and load weight (four levels). Each load weight was carried at each carrying mode, producing a total of eight experimental conditions. The carrying modes were:

(i) Carrying the tool bag on one shoulder

(ii) Carrying the tool bag in the hand

The load weights were:

(i) No tool bag (zero weight)

(ii) 5% of body weight carried

(iii) 10% of body weight carried

(iv) 15% of body weight carried

Most of the studies that were discussed in the literature review section in Chapter 1 conducted their studies using 5%, 10%, 15%, 20%, 30% of the participants' bodyweight as a load to be carried. Since the Institutional Review Board of UTA advised no more than 15% of body weight to be carried on one shoulder, the maximum weight limit that used in this study was 15%.

Specific objectives of the study were to measure and analyze the following:

- Lumbosacral joint (L5-S1) disc compression and shear forces
- pelvis obliquity, tilt, and rotation
- knee joint compression force
- Hip compression force

The University of Texas (UTA's) Institutional Review Board (IRB) approved the protocol for this experimental laboratory study.

3.1 BODY SEGMENT INERTIA PARAMETERS

Quantitative biomechanical analyses involving human motion require estimations of body segment inertia parameters (BSIP). Relevant BSIPs include mass, moment of inertia, location of the center of mass and radius of gyration (Leva, 1996). Clauster et al. obtained mean BSIPs by measuring cadavers of elderly males and those parameters were used widely to estimate initial characteristics of the subjects (Clauser et al., 1969) . However later, it was showed that the generalization of that data could result in significant error while calculating the location of the body centers of mass (de Leva, 1994). De Leva found another mean BSIPs (Zatsiorsky et al. 1990a) that were reliably generalizable to college athletes, where the previously reported errors (de Leva, 1994) were reduced. To locate the center of mass of a particular segment and to define the length of that segment, Zatsiorsky used bony landmarks as reference point. Since several of the reference points were substantially distant from the centers of the adjacent joints, it was difficult to accurately position the segment center of mass while subject performed a dynamic motion like flexion of his/her joint (Leva, 1996). Later, mean relative center of mass and radii of gyration were adjusted by De Leva, 1996 and he used locations of joint centers as reference points instead of bony landmarks.

For defining the segment length, the following equations were used in the study of De Leva, 1996.

$$\bar{l} = \bar{r}_{abs}/\bar{r}_{rel}$$

$\bar{l}$ = mean length of a segment

$\bar{r}_{abs}$ = mean absolute radius of gyration of the respective segment with respect to a particular axis

$\bar{r}_{rel}$ = mean ratio of radius of gyration and length

The values of $\bar{r}_{rel}$ are identical for all the participants (Zatsiorsky et al., 1990)

Mean absolute radius of gyration of a segment about a particular axis could be found using the following equation

$$\bar{r}_{abs} = \sqrt{\bar{I}/\bar{m}}$$

$\bar{I}$ = mean moment of inertia of the segment about the respective axis

$\bar{m}$ = mean segment mass

Hip Joint center locations were obtained using Bennet ,2016.

3.2 TASKS (ACTIVITIES)

To properly maintain the integrity of the experiment, the descriptions of the activities to be performed by the participant were standardized. Because of the complications of attaching sensors to the body and the long duration associated with subject preparation, the eight experimental tests were not performed in a random order but in a pre-defined sequence.

The environment of the experiment (the walking distance, the tool bag, and the position of the force plates) were kept identical for all four of the load carriages conditions and participants. Trial tests in the laboratory showed that 97 reflective markers should be attached to a subjects' body, requiring approximately one and a half hour for preparing a subject, and an additional two hours to conduct the experiment

The laboratory task for each participant involved walking short distance, at a self-determined pace, with the tool bag, under eight different load conditions, as described above was recorded by the Vicon system. The Vicon system tracked the reflective markers attached to the body of the participants and captured the motions associated with the trials. More details on collecting the kinetic and kinematic data collection on Vicon has been described in section 3.5.2. The laboratory facilities for this study included two force plates installed in the walkway of the floor. The experimental space is surrounded by 16 VICON cameras to record the motions of the subject while walking. The detailed description of how the VICON system operates is discussed later in this section.

3.3 SUBJECTS

Almost all the occupational personnel who carry electrical tool bag in the industry are over the age of 18 years (median = 41.2 years). In this experiment, since it was not possible to carry out the activities in an actual work environment, we recruited healthy university male subjects to perform the experimental trials. Their average age was 27 years (Range : 26-28 yr) and average weight, 195 lb (range : 165-225 lb). Females were omitted since gender differences or effects were not under investigation.

3.4 APPARATUS

Sixteen MX T40S cameras (4MP resolution 2336 x 1728) connected to Vicon T-Series motion capture system analysis (Vicon Motion Systems Ltd., Denver, CO) were used to collect data during the lifting tasks at a sampling rate of 200Hz (Figure 1). Two ground reaction force plates with a sampling rate of 1,000 Hz using AMTI Optima OPT400600-2000 (Advanced

Mechanical Technology, Inc., Watertown, MA, USA) were used, capturing ground reaction forces (GRF) and moments synchronized with the video motion capture system (Figure 2).

Figure 1: MX T40S Camera used in this study

An electrical and maintenance tool bag (Jackson-Palmer) weighing 4.14lbs was used as the load carriage system (Figure 3). Ninety-Seven reflective markers were attached to the subject's skin over the anatomical landmarks (Appendix 1) using double-sided tape.

Figure 2: AMTI Optima OPT400600-2000 Force Plates

Figure 3: Electrical and Maintenance tool bag used in the study

3.5 EXPERIMENTAL PROCEDURE

3.5.1 Subject Preparation

Reflective markers (Diameter- 4mm) were attached to various segments of the body to help capture the motion of the subject by the VICON cameras. The marker positions and definitions are given in Appendix 1. The markers were placed on the participants' body (Fig 4). Because of their reflectivity the cameras can easily recognize these markers. By recording the position of these markers throughout the experiment, the camera can determine the position of the participant's body. The markers also help the software Nexus (Section 1.3.2) to define the important segments (Head, Thorax, Femur, Tibia etc.) of the subjects. More detailed information on how the reflective markers, cameras and the software work is discussed in section 3.5.2.

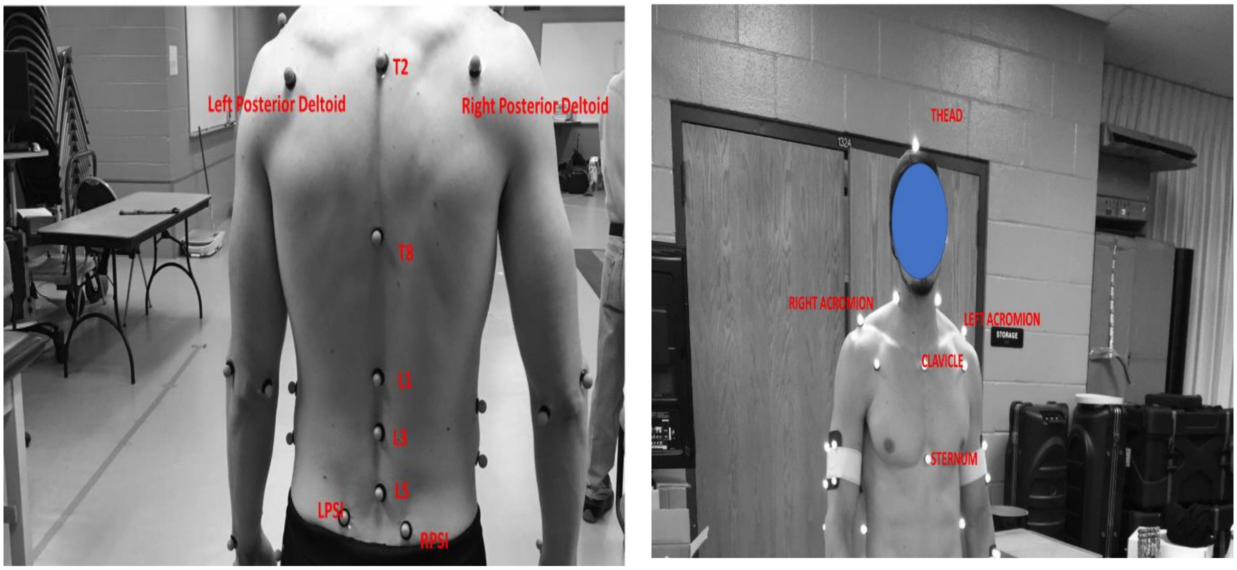

Figure 4 :Attached reflective markers on subject's body to define segments(Torso and Lumbar)

Training for the Subjects: Before the experiment, to make the participants familiarized with the lab environment and the experimental procedures, a brief training session was conducted for each of the participant. Each was briefed about the reflective marker attachment and laboratory cameras that will capture their motions., and about how they will perform their trials with varying weights (Section 3.2). After that, each participant walked along the walkway several times carrying the bag on one shoulder and again in one hand to get himself familiarized with the walkway, the force plates installed on the walkway and the loads.

As mentioned above, each of the subjects was familiarized with the testing conditions and the laboratory environment, and they were given enough time and trials to perform as part of their training to get acquainted with the experiment. With the above marker set attached to the body, each subject walked on the force plate equipped walkway 5 times, each time in a separate weight x carry mode condition, as described earlier.

3.5.2 Kinematic and Kinetic Data Collection on Vicon

The VICON T Series motion capture system was used to perform the motion capture of the participants during the experiment. Once the markers were attached to the participants' body segment, the participant completed three trials without the tool bag. This was done to make the participant familiarize with the laboratory environment and tasks. This familiarization helped the participant to decide on where he should start his trial from so that he could have his step on the right AMTI force plate. The part of the task for the participant was to target the right AMTI force plate, as the study intended to gather information from the force plates which facilitated in marking the stance phase during the time of data processing . That is why several pilot trials were done before moving forward with main experimental trials. The familiarization made sure that aiming for the right force plate would not affect the normal gait of the participant.

To make the VICON system ready to capture the motions of the reflective markers as the participant moved, the cameras needed to be calibrated, with a Vicon calibration device -- an Active wand. This T-shaped calibration wand consists of five LEDs and the wand was set to strobe mode. Then one person waved the wand facing the LEDs towards the camera around the capture volume so that the cameras could capture those five LEDs. As soon as the cameras could see all the five LEDs on the wand, a blue LED began flashing on the strobe. The associated software (VICON NEXUS) confirmed when the calibration was done properly for all the cameras. Then the wand was kept on the force plate to set the volume origin (global coordinate system). This directed the VICON system where the center of the capture volume was and what its orientation (x,y, and z axis) was. In this experiment, the T- shaped active wand, with the LED lights turned on, was placed flat on the middle of the force plates.

After calibration and setting volume origin was done, a static trial was conducted. As part of the trial the subject was instructed to stand still for five to six seconds (Figure 5 & Figure 6). This let the VICON cameras to identify the reflective markers on the participant's body. The tracking helped the VICON NEXUS 2.5 software to create a 3D model of the participant.

Then each subject was asked to perform a series of movements that exercised all his joints for a functional trial. In this, case each subject performed flexion, extension, abduction, adduction of both of their legs and arms, inversions and aversions of ankles, and internal and external rotation of the hip, and a jump at the end.

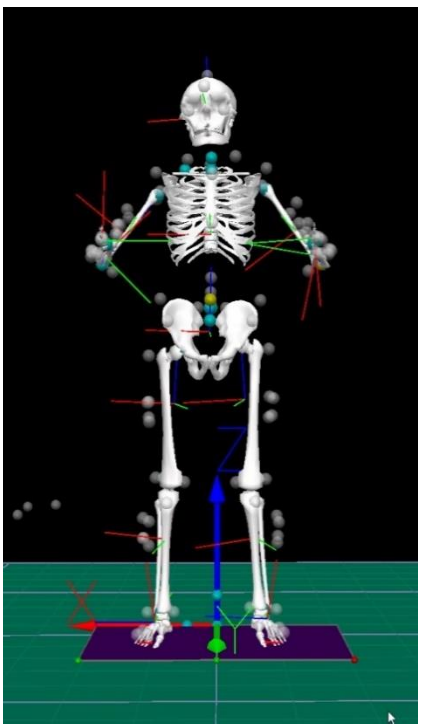

Figure 5: Full Body Frontal View of Reflective Markers Model

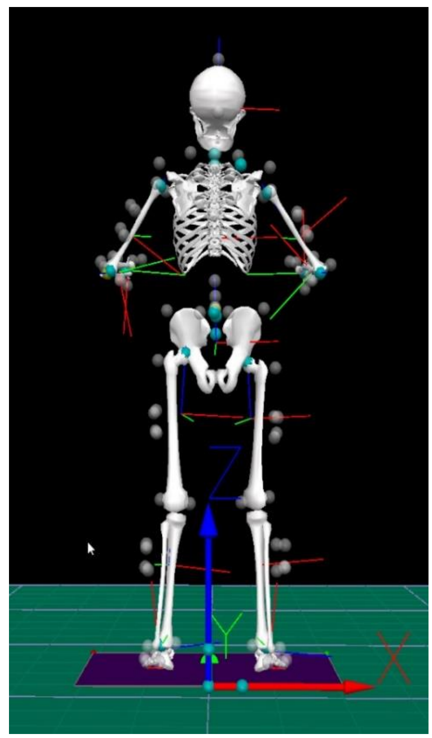

Figure 6: Full Body Rear View of Reflective Markers Model

This functional trial was done to ensure that each joint moves through a range that represents what the subject was assigned to do during the capture of trial data and to get the best results out of the Vicon system. After calibration, the subjects performed the trials under each of the eight conditions stated earlier.

The system recorded the movement of the participants by the video cameras installed in the laboratory for the reproduction of a 3-D image digitally. The system comprised 16 high-resolution VICON cameras. LED (Light Emitting Diodes) strobes surrounded the camera lenses. Then the reflective markers were attached to different segments of the body. As the participant moved, the body segments with the reflective markers were also in motion. Camera lens captured the reflection of the LED lights. A light-sensitive plate was struck by those lights which in turn created a video signal. A software (Vicon Nexus), in association with the motion capture system, was used to collect the original video data. Then the data was processed to reproduce the equivalent digital motion of each segment with respect to the three-dimensional laboratory coordinate system (Figure 7). That helped in obtaining the marker movements in 3D space.

However, there were some frames in the beginning of each video where the whole marker set was not visible. Therefore, the first few frames were eliminated up until the point where all the markers were visible in the reconstructed data so that the software can identify all the attached markers. Even after eliminating the first few frames, there were some frames in the reconstructed data where several markers were missing. To fix this issue, different default functions associated with the software were used to fill those gaps created by the missing markers. After that, the "Visual 3D" (C-Motion, Germantown, MD, USA) software was used to work with the processed data from the Nexus to extract kinetics and kinematics data of different joints and segments we aimed to evaluate. The data was transferred to Visual 3D software format to estimate compression and shear forces on the lumbar spine. As mentioned earlier during the trials, the subject's reflective markers were not always visible and there were unnecessary noises. Marker trajectories were used to filter with a fourth order recursive Butterworth low-pass filter with cutoff frequency of 6 Hz. Ground reaction force data was filtered using a 20th order critically damped filter with cutoff frequency of 30Hz. Three-dimensional joint angles were calculated using an x (flexion/extension), y (abduction/adduction), z (axial rotation) Cardan rotation sequence (Cole et al., 1993). To estimate the internal forces, a fifteen-link segment model was used (Bush-Joseph et al., 1988). The link segments used were the, thorax, neck, both the upper arms, forearms, hands, leg, pelvis, thigh, and foot. For this model, inverse dynamics was used to calculate the reaction forces and moments at particular joints (Bush-Joseph et al., 1988; Chaffin, 1973; De Looze et al., 1994; Freivalds et al., 1984; Lavender et al., 1999). Body segment parameters (mass, CoM location, and moment of inertia) were used based on the findings of Leva (1996) and hip joint center locations were obtained using Bennett (2016) (Bennett et al., 2016; Leva, 1996).

3.6 Laboratory Co-ordinate System

The global coordinate system for our laboratory system was created as described in section 2.8.2. The orientation of the three axes is shown in Figure 7

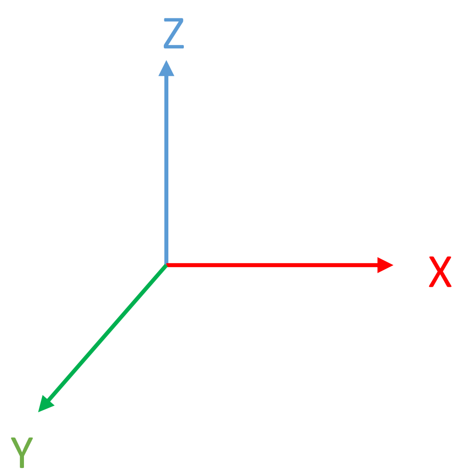

Figure 7 Laboratory Coordinate System for this experiment

The kinetics and kinematics variables along the three axes and their direction are given in the following table

Table 1: The directions of the Kinetics and Kinematics variables with respect to the laboratory coordinate system

Pelvis Relative to Thoracic Segment	**Axis**	**Positive Direction**
Pelvic Obliquity	Y	Ipsilateral
Pelvic Tilt	X	Anterior Tilt
Pelvic Rotation	Z	Internal
Knee Joint Forces	**Axis**	**Positive Direction**
Mediolateral Force	X	Flexion
Anteroposterior Force	Y	Adduction
Compression Force	Z	Internal
L5/S1 Forces	**Axis**	**Positive Direction**
Compression Force	Z	Vertical GRF
Anteroposterior GRF	Y	Anterior GRF
Mediolateral GRF	X	Lateral GRF

3.7 L5-S1 Technical Co-ordinate System

The location of the L5-S1 disc was estimated based upon the typical size of an adult male lumbar vertebrae (Nissan & Gilad, 1986; Zhou et al., 2000). A virtual marker was created in Visual 3D to identify the location of the L5/S1 disc. A technical coordinate system was created to define the orientation of the L5-S1 disc as follows: a unit vector defining the X axis was created by subtracting the LPS marker position from the RPS marker, a unit vector defining the Z axis was created by subtracting LV5 marker position from LV3 position, the Y axis unit vector was defined the cross product of Z unit vector crossed into the X unit vector. The Visual 3D waist force was then transformed into the L5-S1 coordinate system.

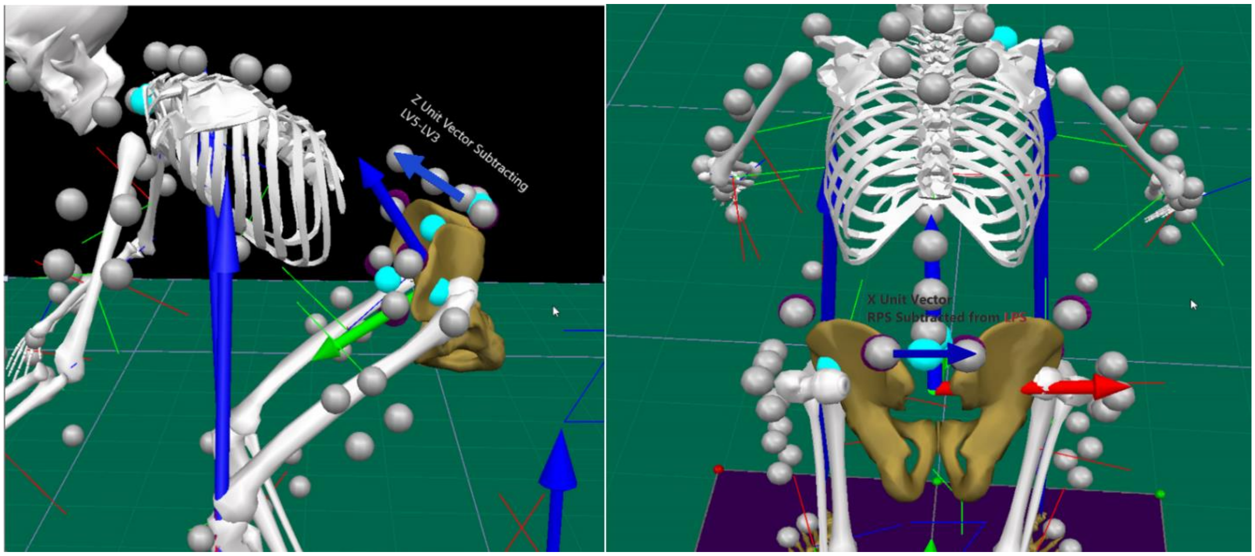

Figure 8 : Technical L5/S1 Coordinate System

3.8 ESTIMATING L5/S1 COMPRESSION FORCE

The force acting on the superior surface of the pelvis (F Pelvis) was transformed into L5/S1 coordinate system. A lever arm from the L5/S1 joint center to the LPP marker was calculated. After that, the waist moment about the Y axis, trunk lateral flexion moment of the coordinate system was divided by the lever arm to get the component due to the action of muscles across the abdominal joint. This component represents the component of compression force required to tilt the upper body when holding the tool bag. The lumbar compression force was computed as the sum of the transformed F Pelvis and F Muscle acting in the L5/S1 coordinate system.

The medial/lateral shear force acting on the L5/S1 system was the X component of the Lumbar force.

$$F_{ab\_com\_force} = \frac{M_y}{r_y}$$

$$F_{L5/S1\_comp} = F_{pelvis\_transformed} + F_{ab\_com\_force}$$

$F_{pelvis\_transformed}$, Force on the superior surface of the pelvis transformed into L5/S1

$M_{y,}$ Trunk Lateral Flexion moment

$r_{y,}$ lever arm from the L5/S1 joint center t o the LPP marker

$F_{ab\_com\_force,}$ force component due to the action of muscles across the abdominal joint

$F_{L5/S1\_comp,}$ L5/S1 compression force.

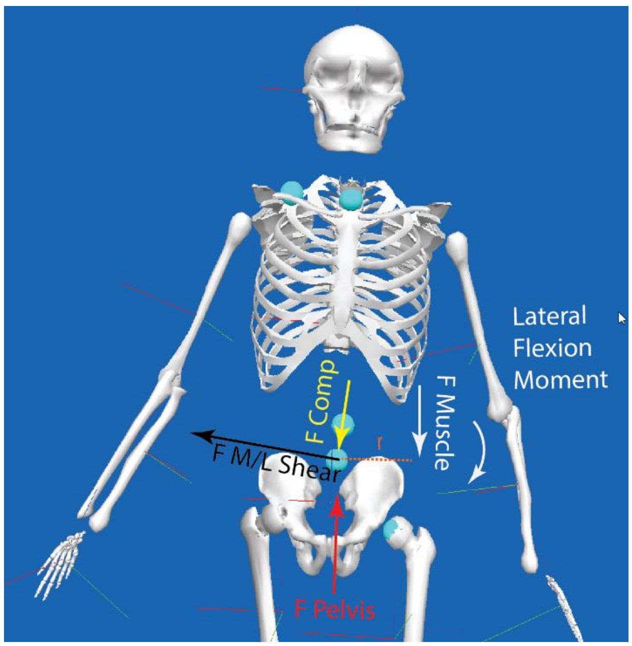

Figure 9: Lumbar Compression Force and MedioLateral Shear Force

3.9 ESTIMATING SHEAR FORCES AT L5/S1

The shear forces at the L5/S1 disc were estimated using visual 3d c-motion software. The anterior-posterior and the medial-lateral shear forces acting on the pelvis were transformed into the lumbar coordinate system at L5/S1 estimated disc location. The anterior-posterior shear force acting on the L5/S1 system was the Y component of the lumbar force. The medial-lateral force acting on the L5/S1 system was in the X component of the lumbar force.

Pelvic tilt, obliquity and rotation, trunk segment angles, knee and hip joint compression forces were also estimated using visual 3d c-motion software.

4 CHAPTER 4: RESULTS

4.1 PEAK COMPRESSION FORCE AT L5/S1

Peak compression force at L5/S1 were estimated according to the methodology described above using Visual 3d c-motion software. The peak compression forces at L5/S1 for one subject while carrying the tool bag on one shoulder under 4 different weight condition (0%, 5%, 10% and 15% of bodyweight) over three trials are graphed in Figure 10.

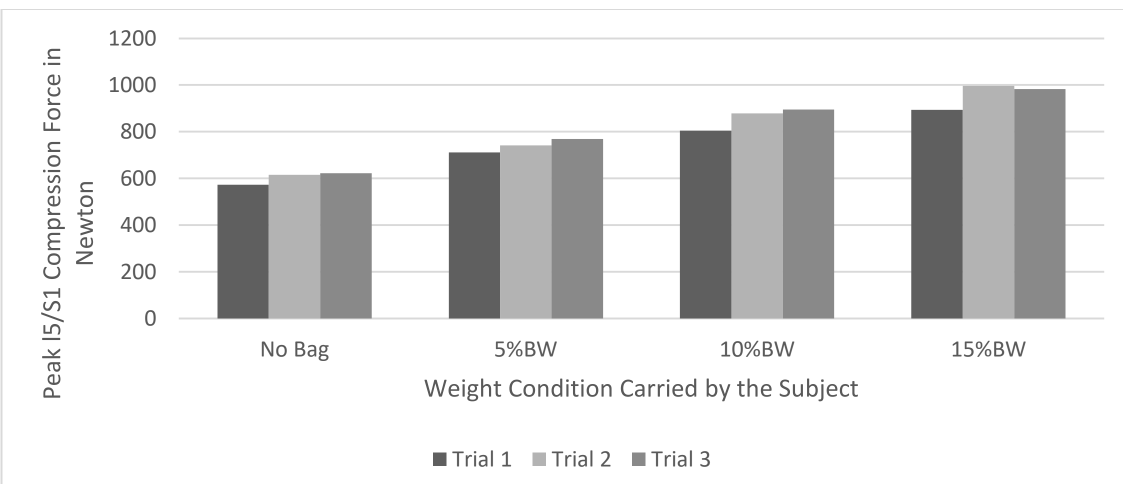

Figure 10: Peak compression forces at L5/S1 disc (from heel strike to toe off) while walking under varying weight carrying conditions (no tool bag, 5%, 10% and 15% of BW) over three trials.

The peak compression forces were calculated during the time between heel strike to toe off. As seen in Figure 10 expectedly, carrying and walking with 15%BW load on one shoulder yielded the highest peak compression force on L5/S1. However, the percentage increase in the average peak compression force tends to decrease as the load carried by the participant increased. From no bag condition to 5% of body weight condition there was an 23% increase in the average peak compression force. The increase was 16% from 5%BW condition to 10%BW condition, and 11% from 10%BW condition to 15%BW condition (Figure 11).

Figure 11: Percentage increase in Average Peak Compression force as the weight carried by the participant increased

4.2 PEAK SHEAR FORCE AT L5/S1

The peak L5/S1 shear forces were calculated from heel strike to toe off using the visual 3D c-motion software. The peak L5/S1 shear force increased as the weight carried by the participant increased (Figure 12). The highest peak shear force at L5/S1 occurred while carrying and walking with 15% BW. The percentage increase in average peak shear force tends to decrease as the weight carried by the participant increased (Figure 13).

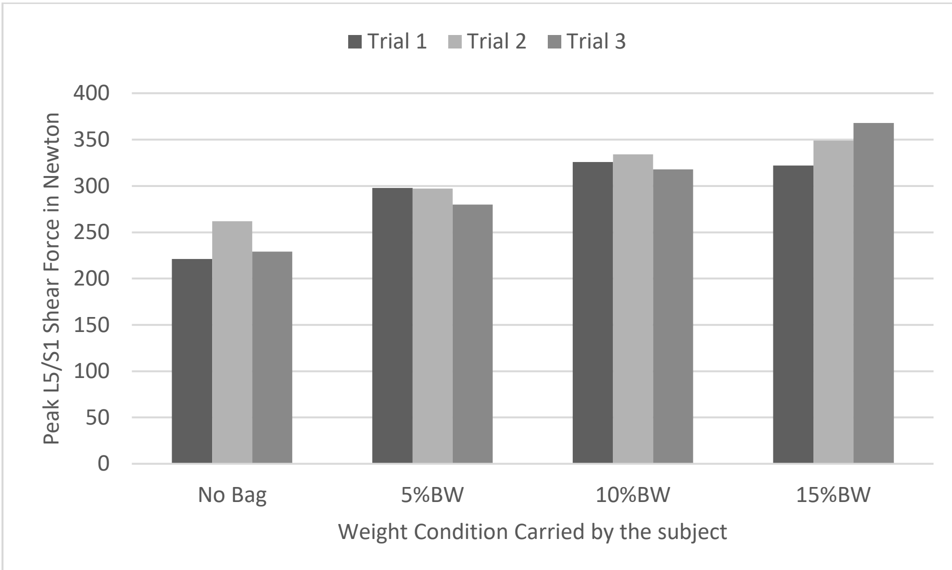

Figure 12 : Peak shear forces at L5/S1 disc (from heel strike to toe off) while walking under varying weight carrying conditions (no tool bag, 5%, 10% and 15% of BW) over three trials.

4.3 PEAK KNEE JOINT COMPRESSION FORCE

The peak knee compression force from heel strike to toe off for one subject while carrying the tool bag under varying weight condition over the three trials are graphed in Figure 14. The highest peak compression force (1138N) was observed during carrying and walking with 15%BW. However, there was a 6% increase from no bag to 5%BW condition, 4% increase from 5%BW to 10%BW and ~~an~~ only 1% increase from 10%BW to 15%BW condition.

4.4 PEAK HIP JOINT COMPRESSION FORCE

The peak hip joint compression forces from heel strike to toe off were calculated using visual 3d c-motion software. The highest peak hip joint compression force (1089N) was observed at 15%BW condition. There was a 9% increase in compression force from no bag to 5%BW condition. However, the percentage increase (4%) from 5%BW to 10%BW and from 10%BW to 15%BW remained the same.

Figure 13:Percentage increase in Average Peak Shear force at L5/S1 as the weight carried by the participant increased

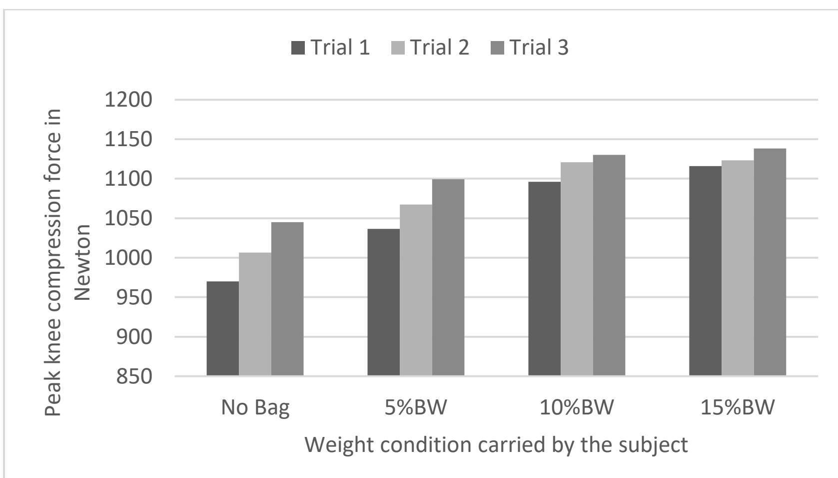

Figure 14: Peak Knee Joint Compression forces (from heel strike to toe off) while walking under varying weight carrying conditions (no tool bag, 5%, 10% and 15% of BW) over three trials.

4.5 PELVIC ANGULAR MOTION

The mean angular pelvic tilt tends to decrease during the heel strike to toe off as the weight carried by the participant increased (Figure 16). However, for this participant, the deviation of mean pelvic tilt at 5% BW,

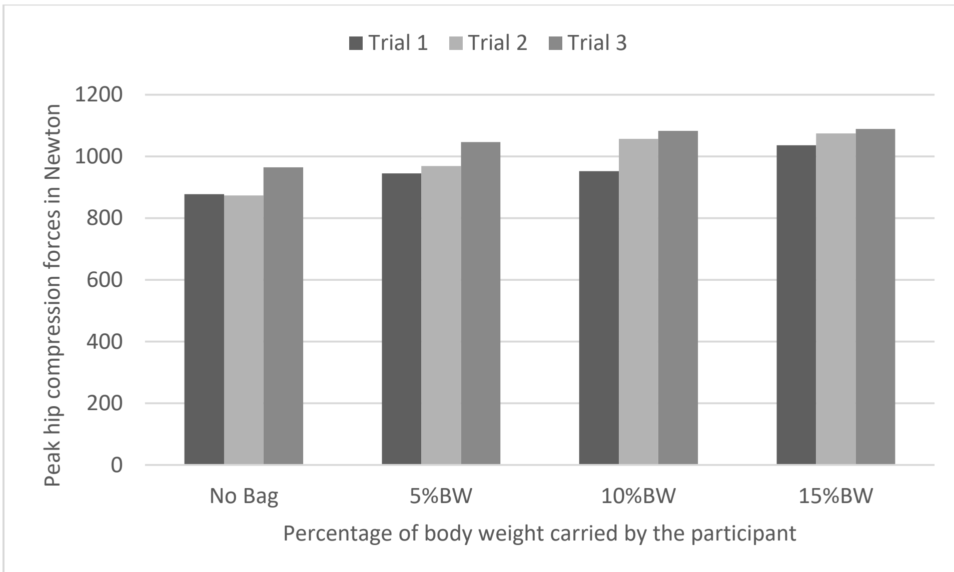

Figure 15:Peak Hip Joint Compression forces (from heel strike to toe off) while walking under varying weight carrying conditions (no tool bag, 5%, 10% and 15% of BW) over three trials.

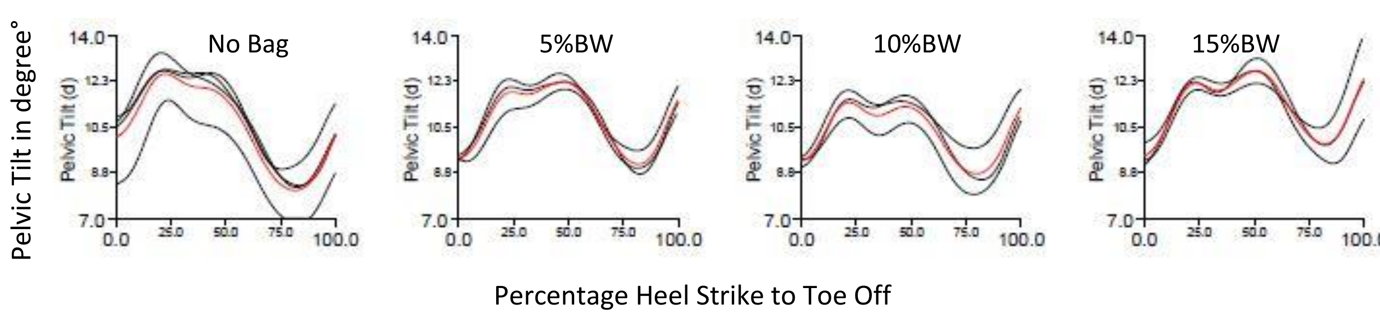

Figure 16: Changes in angular pelvic tilt during heel strike to toe off under varying weight condition (No bag, 5%BW, 10%BW, and, 15%BW)

10%BW, and 15%BW conditions compared to the no bag condition was very little and the result was similar with other studies. Mean angular pelvic rotation during the heel strike to toe off did not change much apparently for the participant (Figure 17). Pelvic obliquity increased under all the three weight conditions (5%, 10% and 15% BW) compared to walking without the tool bag (Figure 18).

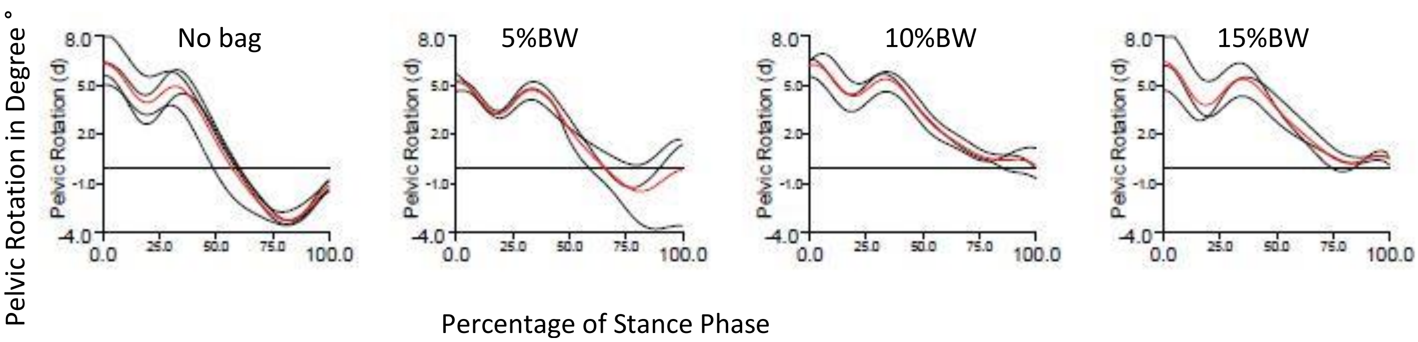

Figure 17: Changes in angular pelvic rotation during heel strike to toe off under varying weight condition (No bag, 5%BW, 10%BW, and, 15%BW)

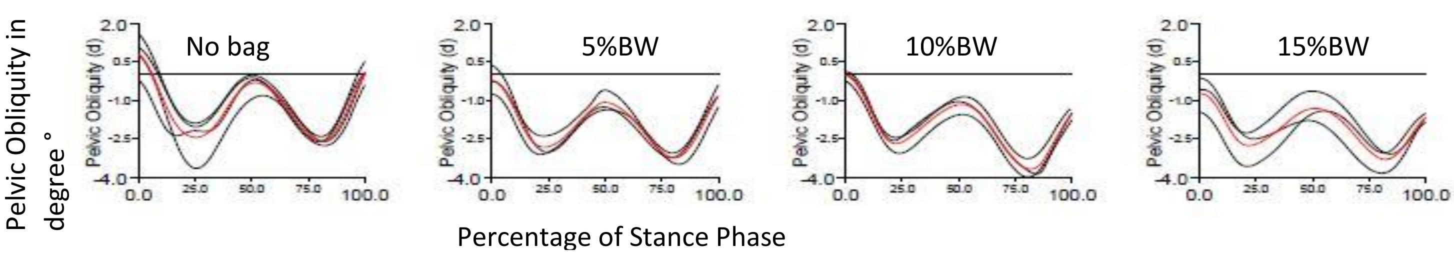

Figure 18: Changes in angular pelvic obliquity during heel strike to toe off under varying weight condition (No bag, 5%BW, 10%B, and 15%BW)

5 CHAPTER FIVE: DISCUSSION

The novelty of this study was to create a technical coordinate system, as part of the methodology, that can precisely locate the L5-S1 joint center and measure the force distribution precisely there. In this study, the waist force calculated from the Visual 3D was transformed to that newly created technical co-ordinate system to measure the precise joint forces at the L5-S1 joint. Locating the L5-S1 joint center as precisely as possible helps in predicting the joint forces acting at the L5-S1 joint more accurately. Generally, past studies used the data from the body motion of the nodal joints as part of their model for the calculation of the spinal load. The anatomical marker at the L5-S1 level position was considered as the L5-S1 joint center for several studies (Cheng et al., 1998; Potvin, 1997). This means the joint center was considered located at the externally attached marker positions.

Several studies used the mean value from the CT scan to locate the lumbosacral joint center in their biomechanical model (Goh et al., 1998; Khoo et al., 1995). In a study, a calibrated lumbosacral joint center was located based on static markers at the PSIS (Posterior Superior Illiac Spine)(Li et al., 2019). Their adaption of locating the L5-S1 joint center was considering the L5-S1 joint to be a joint on the bony pelvis (Reed et al., 1999). However, Reed et al., 1999 assumed this for studies that require participants to be in the seated posture.

However, in this study, the externally located anatomical marker attached to the skin at L5 level was not considered as the L5-S1 joint center. This current study developed a new measure to precisely locate the L5-S1 joint center co-ordinate system by transferring the waist coordinate system to the newly created technical coordinate system at the L5/S1 disc.

Moreover, the participants of most of the studies discussed in the literature review section carried their load carriage system above the T8 (eighth vertebrae of the thoracic region) level, whereas in this study the participant walked while carrying the tool bag below the T12 (12^{th} vertebrae of the thoracic region) level. This condition also had its effect on the results of this study that was not the same as the results of other similar studies.

The changes in the average peak compression force at the L5/S1 disc found in this study were consistent with other previous studies. The percentage increase (over 25%) in average peak compression at L5/S1 from no bag condition to 15%BW condition was found to be similar to the force found by Goh et al., 1998. In the present study, the increase in average peak lumbosacral force was not of same proportion as the increase in the tool bag load. The percentage change in average peak compression force from 5%BW to 10%BW and 10%BW to 15%BW decreased, due to the increased knee flexion as the weight increased from 5%BW condition. The participant compensated the extra load by flexing his knee more as the weight increased. Like Goh et. al, 1998, the peak lumbosacral compression force was comprised mostly of the compressive load acting externally at the L5/S1, with much lower shear forces.

The knee and hip compression forces increased as the weight carried by the participant increased. This might be due to the increased knee and hip adduction moment (Hall et al., 2013)

Pelvic tilt and rotation seemed to have a very little change as the weights were added to the shoulder which was consistent with a previous study (Smith et al., 2006). However, in contrast to Smith et al., the current study produced an increase pelvic obliquity. The reason might be the added weight inside the tool bag suspended from the right shoulder caused the right hip to raise above the left hip.

One of the reasons for greater increase in peak compression force at L5/S1 as the weight carried by the participant increased might be the increased lateral bending of the trunk (Figure 19) and the backward trunk bending. (Figure 20). Forward lean of the trunk during heavy load carriage has previously been reported by several researchers. (Harman et al., 2000; Kinoshita, 1985). Due to the heavy weight the participant might try to compensate the hip moments by forwarding the trunk lean which may result in increased lordosis and might contribute to the compression of the lumbar vertical bodies. Increased forward lean while carrying the loaded tool bag is necessary to stabilize the shifted center of mass (Majumdar et al., 2010). In this study a backward bending of the body with 5%, 10%, and 15% BW compared to no bag condition was observed. Previous research also confirmed that even with a little increase of the load at the back can cause forward lean (Grimmer et al., 2002) and backward lean in some cases (Singh & Koh, 2009).

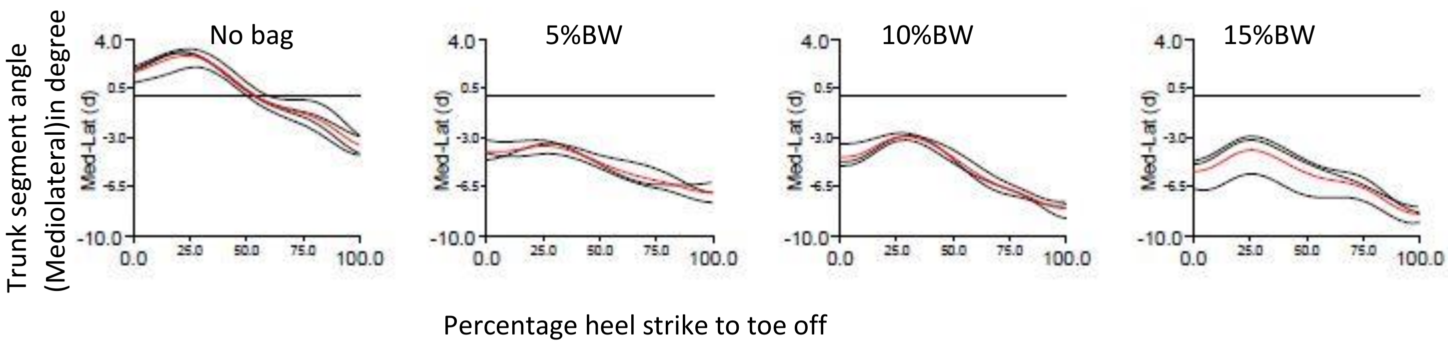

Figure 19:Changes in mediolateral trunk angle during heel strike to toe off under varying weight conditions

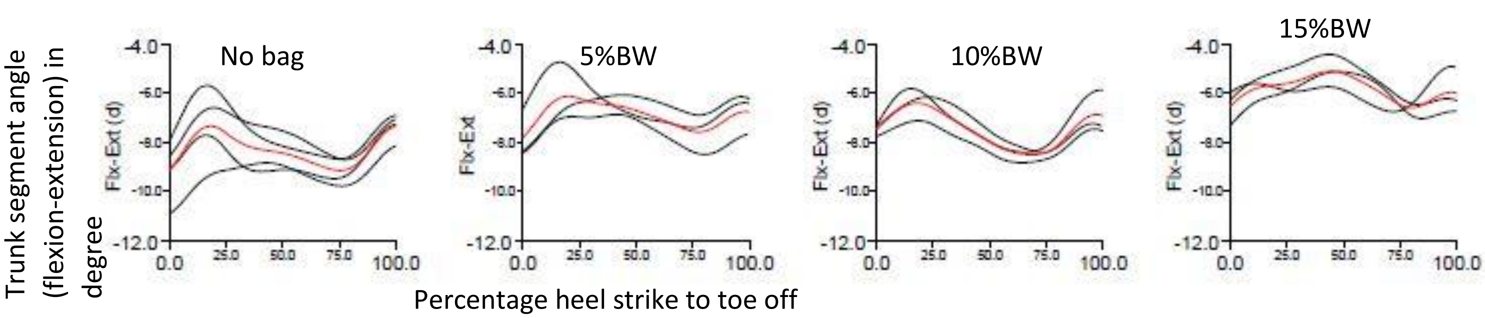

Figure 20: Changes in flexion-extension trunk angle during heel strike to toe off under varying weight conditions

The studies found increased forward lean during load carriage also considered that wearing a load carriage system at the upper trunk (above T8-T9) level would result in greater forward lean.

However, in this present study, it was observed that the participant carried a tool bag at below the T12-L1 level ,weighing 5%, 10% and 15% of his bodyweight and walked, he tried to stabilize the center of mass of the system (Body and Tool Bag) compared to the no bag condition resulting in more backward lean. Therefore, the participant had a backward movement to achieve the adjustment that helped the body in minimizing the energy expenditure. The increase in trunk backward movement is consistent with other studies that positioned the load carriage system below the upper trunk level (Al-Khabbaz et al., 2010).

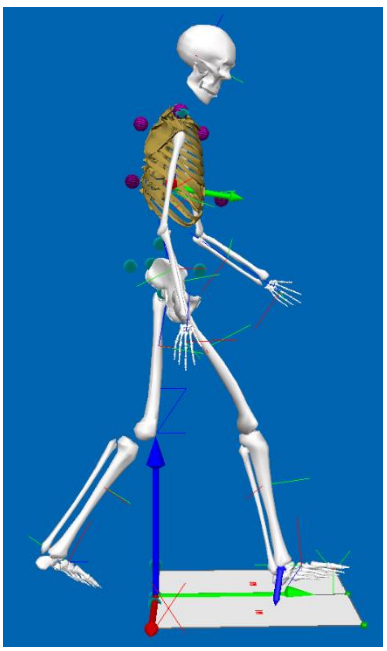

Figure 21:No Tool bag at Heel strike

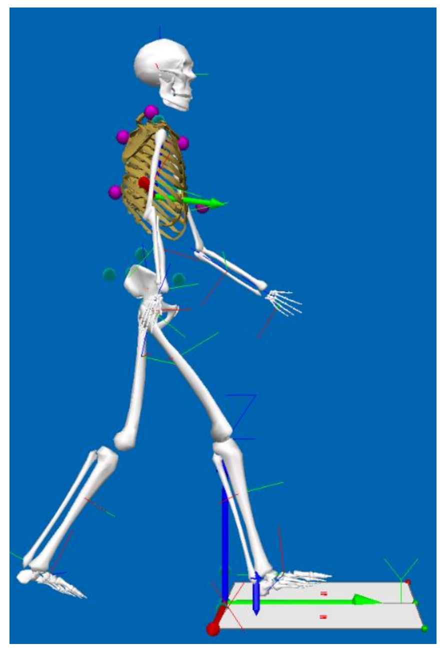

Figure 22: Heel strike at 15%BW condition

The deviation from natural posture due to the unilateral carriage of the tool bag causes low back strain (Chaffin & Andersson, 1991). (Goh et al., 1998) also found that the effective progression of the trunk forward and backward might result in peak lumbosacral forces when subject carried load posteriorly. In this study, from heel strike to toe off, it was observed that the peak compression force at lumbosacral joint increased as the weight condition carried by the participant increased. The trunk extension or backward inclination also increased (Figure 20, 21 & 22) as the weight increased on the back which contributed to that increased peak compression force at the L5/S1 joint. Maintaining excessive forward inclination or backward inclination for longer period may cause fatigue and increased stress upon back muscles and discs.

From Figure 9, shoulder, and trunk asymmetry due to unilateral carriage was also observed. This asymmetry is a compensatory position for placing too much weight on the participant's shoulder. Trunk lateral flexion was observed (Figure 19) and the change in the trunk lateral flexion was apparently more than the trunk extension as weight condition carried by the participant increased. The trunk shifted in the opposite direction (left) to the shoulder from which the tool bag was suspended (right shoulder), due to lateral flexion of the spine towards the right side, as a compensatory mechanism. (Drzał-Grabiec et al., 2015). Other studies(Akbari & Gannad, 2006; P. et al., 2004) have also found increased asymmetry in trunk while carrying load unilaterally.

The forward bending and the lateral flexion of the trunk contributed to increased spinal bending torque. Therefore, carrying a tool bag asymmetrically results in increased spinal overload and affects the lumbosacral joint adversely.

6 CHAPTER 6: CONCLUSION

In this study we tried to develop a new methodology, for the estimation of forces at a more specific location of L5-S1 disc, for carrying a load on the body asymmetrically. A technical lumbar coordinate system was created to define the location of the L5-S1 disc. An electrical and maintenance tool bag was used as the load carriage system. The tool bag tended to sway side to side which disturbed the normal gait cycle of the participant. To prevent the swaying, the subject needed to hold the bag with the hand associated with the shoulder over which the tool bag was suspended from creating restriction of the arm. We were able to estimate the peak compression force and shear force at L5-S1 disc under varying weight condition carried by the participant while walking. In addition to that, we also estimated the knee and hip joint compression forces and the angular changes in pelvis (tilt, obliquity and rotation) with respect to the three axes.

The study was supposed to recruit more participants for the data collection but unfortunately due to the ongoing pandemic of Covid-19 the data collection was affected since the data collection required human interactions. We, therefore, worked with the data from one subject that was collected as part of the pilot study. Another limitation we faced was the suspension of the tool bag over the shoulder. The tool bag tended to hide the thigh cluster markers and the cameras found it difficult to process those clustered markers data adequately for defining the thigh segment. The positioning of the markers to the exact anatomical position was one of the challenges. Any movement of the marker could result in missing markers in the software which disturbs the data processing and generation of the link segment model. However, in this study, the method developed for locating the L5-S1 joint coordinating system precisely for measuring the L5-S1 disc compression and shear force was able to estimate the desired forces at L5/S1 disc.

The method developed in this study can be used in future studies related to lifting and carrying heavy loads in industrial sectors. For instance, further research can be done using this methodology to compare different asymmetrical carrying conditions during ground walking or stair ascend or descend by simulating the task environment in a laboratory within the area covered by the Vicon cameras.

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8 APPENDIX A

8.1 REFLECTIVE MARKERS SET ON DIFFERENT BODY SEGMENT

Upper Body

Head

- THead = Top of Head
- PHead = Posterior Head
- C7 = Cervical Vertebrae 7
- Ahead = Anterior Head
- RHead = Right Head
- LHead = Left Head
- RNeck = Right Neck
- LNeck = Left Neck

Trunk

- RAC = Right Acromial Joint
- LAC = Left Acromial Joint
- CLAV = Clavicle
- STERN = Sternum
- T2 = Thoracic Vertebrae 2
- T8 = Thoracic Vertebrae 8
- RLT = Right Lower Trunk (lowest floating rib on right side)
- LLT = Left Lower Trunk (lowest floating rib on left side)

- RUT = Right Upper Trunk (in line with the base of the sternum)
- LUT = Left Upper Trunk (in line with the base of the sternum)

Lumbar

- L1 = First Lumbar Vertebrae
- L3 = Third Lumbar Vertebrae
- L5= Fifth Lumbar Vertebrae

Right Upper Arm

- RADL = Right Anterior Deltoid
- RPDL = Right Posterior Deltoid
- RUA1 = Right Upper Arm 1 (cluster)
- RUA2 = Right Upper Arm 2 (cluster)
- RUA3 = Right Upper Arm 3 (cluster)
- RUA4 = Right Upper Arm 4 (cluster)
- RLEL = Right Lateral Elbow

Right Forearm

- RMEL = Right Medial Elbow
- RFA1 = Right Forearm 1 (cluster)
- RFA2 = Right Forearm 2 (cluster)
- RFA3 = Right Forearm 3 (cluster)
- RFA4 = Right Forearm 4 (cluster)

- RWRR = Right Wrist Radial
- RWRU = Right Wrist Ulnar

Right Hand

- RHR = Right Hand Radial
- RHM = Right Hand Middle
- RHU = Right Hand Ulnar

Left Upper Arm

- LADL = Left Anterior Deltoid
- LPDL = Left Posterior Deltoid
- LUA1 = Left Upper Arm 1 (cluster)
- LUA2 = Left Upper Arm 2 (cluster)
- LUA3 = Left Upper Arm 3 (cluster)
- LUA4 = Left Upper Arm 4 (cluster)
- LLEL = Left Lateral Elbow

Left Forearm

- LMEL = Left Medial Elbow
- LFA1 = Left Forearm 1 (cluster)
- LFA2 = Left Forearm 2 (cluster)
- LFA3 = Left Forearm 3 (cluster)
- LFA4 = Left Forearm 4 (cluster)
- LWRR = Left Wrist Radial
- LWRU = Left Wrist Ulnar

Left Hand

- LHR = Left Hand Radial
- LHM = Left Hand Middle
- LHU = Left Hand Ulnar

Lower Body

Pelvis

- RPP = Right Pelvis Peak
- LPP = Left Pelvis Peak
- RPS = Right PSIS (Posterior Superior Iliac Spine)
- LPS = Left PSIS (Posterior Superior Iliac Spine)

Right Thigh

- RHP = Right Hip (Greater Trochanter)
- RTH1 = Right Thigh 1 (cluster)
- RTH2 = Right Thigh 2 (cluster)
- RTH3 = Right Thigh 3 (cluster)
- RTH4 = Right Thigh 4 (cluster)
- RLK = Right Lateral Knee
- RMK = Right Medial Knee

Right Shank (lower leg)

- RTT = Right Tibial Tuberosity
- RSK1 = Right Shank 1 (cluster)
- RSK2 = Right Shank 2 (cluster)
- RSK3 = Right Shank 3 (cluster)

- RSK4 = Right Shank 4 (cluster)
- RLA = Right Lateral Ankle
- RMA = Right Medial Ankle

Right Foot

- R1MH = Right 1st Metatarsal Head
- RToe = Right Toe (in between 2nd and 3rd metatarsal head)
- R5MH = Right 5th Metatarsal Head
- R5MB = Right 5th Metatarsal Base
- RHL = Right Heel

Left Thigh

- LHP = Left Hip (Greater Trochanter)
- LTH1 = Left Thigh 1 (cluster)
- LTH2 = Left Thigh 2 (cluster)
- LTH3 = Left Thigh 3 (cluster)
- LTH4 = Left Thigh 4 (cluster)
- LLK = Left Lateral Knee
- LMK = Left Medial Knee

Left Shank (lower leg)

- LTT = Left Tibial Tuberosity
- LSK1 = Left Shank 1 (cluster)
- LSK2 = Left Shank 2 (cluster)
- LSK3 = Left Shank 3 (cluster)
- LSK4 = Left Shank 4 (cluster)

- LLA = Left Lateral Ankle
- LMA = Left Medial Ankle

Left Foot

- L1MH = Left 1st Metatarsal Head
- LToe = Left Toe (in between 2nd and 3rd metatarsal head)
- L5MH = Left 5th Metatarsal Head
- L5MB = Left 5th Metatarsal Base.
- LHL = Left Heel