THE ISOLATION OF MUSCLE ACTIVITY AND GROUND REACTION FORCE PATTERNS ASSOCIATED WITH POSTURAL CONTROL IN FOUR LOAD MANIPULATION TASKS

BY

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ABSTRACT

Although much effort has been placed into the reduction of risks associated with manual materials handling, risk of musculoskeletal disorder development remains high. This may be due to the additional muscle activity necessary for the maintenance of postural equilibrium during work tasks. This research proposes that postural control and subsequent additional muscle activity is influenced by the magnitude of the external load and the degree of body movement. The objective of this research was to identify whether performing tasks with increased external load and with a greater degree of trunk motion places additional strain on the musculoskeletal system in excess of that imposed by task demands. Twenty-four male and twenty-four female subjects performed four load manipulation tasks under three loading conditions (0.8kg, 1.6kg, and 4kg). Each task comprised of a static and dynamic condition. For the static condition, subjects maintained a stipulated posture for ten seconds. The dynamic condition required subjects to move and replace a box once every three seconds, such that a complete lift and lower cycle was performed in six seconds. Throughout task completion, muscle activity of six pairs of trunk muscles were analysed using surface electromyography. This was accompanied by data regarding ground reaction forces obtained through the use of a force platform. After the completion of each condition subjects were required to identify and rate body discomfort. Differential analysis was used to isolate the muscle activity and ground reaction forces attributed to increased external load and increased trunk movement.

It was found that the heaviest loading conditions (4kg) resulted in significantly greater ($p<0.05$) muscle activation in the majority of muscles during all tasks investigated. The trend of muscle activity attributed to load was similar in all significantly altered muscles and activation was greatest in the heaviest loading condition. A degree of movement efficiency occurred in some muscles when manipulating loads of 0.8kg and 1.6kg. At greater loads, this did not occur suggesting that heavier loading conditions result in additional strain on the body in excess of that imposed by task demands. In manipulated data, trend of vertical ground reaction forces increased with increased load in all tasks. Sagittal movement of the centre of pressure attributed to load was significantly affected in manipulated data in the second movement phase of

the "hip shoulder" task and the second movement phase of the "hip twist" task. The "hip reach" task was most affected by increased load magnitude as muscle activity attributed to load was significantly different ($p<0.05$) under increased loading conditions in both movement phases in all muscles. Further, a significant interactional effect ($p<0.05$) between condition and data point was found in all muscles with the exception of the right and left lumbar erector spinae during the second movement phase of the "hip reach" task.

Muscle activity associated with increased trunk motion resulted in additional strain on the trunk muscles in the "hip shoulder" and "hip reach" tasks as muscle activity associated with the static component of each of the above tasks was greater than that of the dynamic tasks. Trend of ground reaction forces attributed to increased trunk motion generally increased under increased loading conditions. Additionally, a significant interactional effect ($p<0.05$) between load and muscle activity pattern was found in all muscles during all tasks, with the exception of the right rectus abdominis in the first movement phase of the "hip shoulder' task, the left rectus abdominis in the second movement phase of the "hip knee" task and the right latissimus dorsi during the first movement phase of the "hip twist" task. This was accompanied by a significant interactional effect ($p<0.05$) between load and sagittal centre of pressure movement attributed to load, in both movement phases of all tasks investigated.

From this research it can be proposed that guidelines may underestimate risk and subsequently under predict the strain in tasks performed with greater external loads as well as tasks which require a greater degree of trunk motion. Therefore, this study illustrates the importance of the consideration of the muscle activity necessary to maintain postural equilibrium in overall load analyses.

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LIST OF ABBREVIATIONS

COP	Center of pressure
COM	Center of mass
BOS	Base of support
CNS	Central nervous system
APA	Anticipatory postural adjustment
LBD	Low back disorder
LBP	Low back pain
LES	Lumbar erector spinae muscle
TES	Thoracic erector spinae muscle
EO	External oblique muscle
IO	Internal oblique muscle
RA	Rectus abdominis muscle
LAT	Latissimus dorsi muscle
F_x	Sagittal ground reaction force
F_y	Lateral ground reaction force
F_z	Vertical ground reaction force
COP_x	Movement of the center of pressure in the sagittal plane
COP_y	Movement of the center of pressure in the lateral plane
%MVC	Percent of maximum voluntary contraction

CHAPTER I
INTRODUCTION

BACKGROUND TO THE STUDY

Despite vast developments in automation and mechanisation, manual materials handling (MMH) is still prevalent in industrial settings (Davis and Marras, 2000). Further, despite extensive ergonomics research and interventions, the incidence of injury from manual work is still high (Marras, 2000). Although the question of correct injury surveillance and reporting is valid, that is beyond the scope of this research which questions whether the contribution of postural control to lifting mechanics may have a greater impact on injury risk than previously considered. The most common group of injuries are low back injuries and low back disorders (Snook, 1978; Bigos *et al.,* 1986; Dempsey, 1998; Lavender *et al.,* 2006) and while most research has tended to ignore the impact of postural muscle control, it has been shown fairly recently that the risk of occupationally-related low back disorders may be associated with the stabilisation of the spine (Granata and Wilson, 2001). Thus, the contribution of the trunk muscles to overall trunk muscular load, through maintaining spinal stability and postural equilibrium has not been completely clarified.

Lifting has been identified as one of the most common manual handling tasks (Ciriello *et al.*, 1999) and prolonged exposure to repetitive lifting has been identified as a significant risk factor in the development of musculoskeletal disorders (Marras *et al.,* 2006). Further, repetitive lifting (Frymoyer *et al.,* 1983; Marras *et al.,* 1993; Marras *et al.,* 1995; Marras, 2003), asymmetrical lifting (Marras and Sommerich, 1991; Granata and Marras, 1993) as well as lifting with extended horizontal reach (Ciriello, 2007) have been shown to be especially stressful on the musculoskeletal system. In dynamic tasks, such as lifting, two sets of responses are used to ensure task completion. The first controls ongoing body movement while the second maintains body equilibrium against both internal and external perturbations (Massion, 1992). During lifting, the body's centre of mass (COM) is shifted forward as a result of the external load and this disturbs whole body equilibrium (Toussaint *et al.,* 1997a, b). Lifting tasks require the execution of a flexion movement as well as the control of dynamic balance. Therefore, postural control during lifting activities is essential in order for the task to be completed effectively (Commissaris and Toussaint, 1997; Toussaint *et al.,* 1997b). Despite this, the impact of this control on movement patterns and pattern of muscle activity has largely been ignored.

Despite much effort in recent years being placed on the reduction of the risks associated with MMH, most industries are still seeing a rise in injury profile and while the development of lifting guidelines has proved useful, their efficacy in terms of injury reduction is questionable. This may be explained by the fact that these guidelines do not accurately account for the underlying postural mechanisms which are necessary for safe and efficient manual handling (Kollmitzer *et al.,* 2002). Only a few studies have focused on the postural control strategies which are employed while performing manual handling tasks (Toussaint *et al.,* 1997a; Toussaint *et al.,* 1998; Oddsson *et al.,* 1999). This is despite the fact that successful task completion is dependent on the maintenance of an upright posture which is achieved through the process of postural control (Zedka *et al.,* 1998; Prioli *et al.,* 2006). The trunk musculature performs a dual function as they contribute to the structural stability of the spinal column as well as maintain the body's equilibrium while standing (McGill *et al.,* 2003). The importance of the trunk muscles in providing stability to the lumbar spine is well established and has been confirmed by a number of studies (Cholewicki and McGill, 1996; Cholewicki *et al.,* 1997; Granata and Marras, 2000).

In order to ensure stability and normal functioning of the lumbar spine, postural control is essential in both static and dynamic situations (Silfies *et al.,* 2003). In dynamic conditions, the trunk muscles must be recruited in an appropriate sequence and the strength of the contractions must be great enough to support loads placed on the spine as well as maintain the stability of the spine. Errors in this motor control system may lead to a loss of spinal stability which would result in injury to the lumbar spine (Cholewicki and van Vliet, 2002). It has been established that all trunk muscles contribute to spinal stability but the contribution of each muscle will change under different loading conditions and according to the muscle recruitment patterns employed for task completion (Cholewicki and van Vliet, 2002). Additionally, the role and contribution of each muscle will change throughout task completion (Cholewicki and van Vliet, 2002; McGill *et al.,* 2003). Although the neuromuscular control of spinal stability has been characterised during static tasks, it has received little attention during dynamic movement (Granata and England, 2006). Spine stability can be increased through co-contraction of the agonist and antagonist trunk muscles (Gardner Morse *et al.,* 1995; Cholewicki and McGill, 1996; Cholewicki *et al.,* 1997) but increases in trunk muscle co-activity results in increased spinal loading (Granata and Marras, 1995a; Gardner-Morse and Stokes, 1998) which may impair trunk postural control (Lee *et al.,* 2008).

Postural control is a complex process which involves the integration of sensory information and the generation of appropriate postural responses which are used to either retain, or regain, the body's COM within the base of support (Akram *et al.*, 2008). This allows for upright sitting, stance and gait to be possible and also provides a background postural set necessary for the development and execution of all skilled motor responses (Zedka *et al.,* 1998). To maintain an upright standing position, the central nervous system (CNS) co-ordinates motion across joints and muscles using sensory information gained from the visual, somatosensory and vestibular systems (Chow *et al.,* 1999; Radebold *et al.,* 2001; Akram *et al.,* 2008). As postural coordination and the role of the sensory systems change according to task constraints, the CNS uses different strategies to stabilise the body under different task conditions (Akram *et al.,* 2008). Even though it has been shown that postural control is task dependent (Horak and MacPhearson, 1996) only a few studies have considered the influence of task demands on the mechanisms of postural control (Streepy and Angulo-Kinzler, 2002; Amiridis *et al.,* 2003; Prioli *et al.,* 2006). In addition, investigations regarding the maintenance of equilibrium have tended to focus on the lower extremities and therefore only limited information is available on the behaviour of the trunk during balance and postural control (Preuss and Fung, 2007).

In order to comprehend the effect of external loads on the body, the effect of internal loading must first be fully recognized. Understanding the effect of postural load would aid in a greater awareness of the mechanisms associated with internal loading. This, however, is a complex task as the mechanisms of postural control underly all movement (Houdijk *et al.,* 2009). This study therefore sought to establish the impact of load mass and trunk motion on trunk muscle activity and postural stability.

STATEMENT OF THE PROBLEM

The main problem addressed in this research was to quantify the contribution of the trunk muscles to postural control during the manipulation of different load masses while performing different tasks with varying degrees of trunk motion. It was contended that the additive strain placed on the body, due to postural muscle control under increasing load and different degrees of trunk motion, may further exacerbate overall muscle recruitment patterns, resulting in increased levels of co-contraction. This would ultimately result in decreased spinal tissue tolerance. Thus, in accordance with biomechanical logic, a reduction in tolerance of the spinal tissues due to increased muscle activation may increase the risk of injury during manual handling tasks in which there is a greater need for postural control.

DELIMITATIONS

This investigation was delimited to the responses of 24 male and 24 female subjects who were healthy and free from musculoskeletal injury. In addition, subjects used were university students and subsequently had no manual handling experience. Elite athletes and sedentary individuals were excluded from selection to prevent distortion of data.

Subjects were required to follow a standardised protocol and body posture was controlled by stipulating foot position and by relativising the workstation according to each subject's anthropometric data. Subjects were required to complete four different load manipulation tasks under both static and dynamic conditions. All conditions were completed with three relatively light external loads (0.8kg, 1.6kg and 4kg). Therefore 24 conditions were completed by each subject. Dynamic conditions involved the transfer of a box within designated areas at a rate of one lift and lower every six seconds. The corresponding static position associated with each task was held for ten seconds.

Data regarding muscle activity and ground reaction forces were collected throughout the testing session using a Mega ME6000P16 electromyography unit and a Bertec force platform respectively. The Body Discomfort Map and Scale was used to identify any discomfort experienced during task completion.

LIMITATIONS

The investigation took place in a laboratory setting and therefore environmental factors which may affect workers *in situ* did not affect the results obtained in this investigation. Consequently, the results obtained in this investigation may not be an accurate reflection of the responses in a real world setting due to the difference in environment and working conditions between laboratory and field.

The subjects used for experimentation were university students who had no experience in manual handling and were also not work hardened. As such, responses of experienced individuals may differ from those gathered from this investigation. Subjects may also have differed from the working population in terms of socio-economic status, culture and experience. Although degree of physical activity was controlled, subject lifestyles and hobbies could not be controlled.

The use of surface electromyography may be seen as a limitation in this investigation as it was not possible to measure the activity of all muscles which may contribute to postural stability of the trunk such as the lumbar multifidus and the quadratus lumborum. In addition, the trunk muscles are large in relation to the area of the electrode. Consequently, muscle activation may not be uniform throughout the muscle in which case only a portion of the muscle activation was measured (Masani *et al.,* 2009). Trunk muscles, specifically the oblique muscles, have variable fibre orientation across the muscle which suggests that different portions of the muscle may have different mechanical actions. Multiple electrode sites for the rectus abdominis and external oblique muscles have been suggested by Davidson and Hubley-Kozey (2005). However, this cannot be accomplished with surface analysis of the external oblique muscle and this method also interferes with normalization of the EMG signal while finding MVC values for each portion of the muscle.

Task duration was short with subjects performing each dynamic condition for a period of one minute (ten lifts) and each static position for ten seconds. A minute rest period was also provided between each task. This limits the investigation as work in industrial environments occurs over an eight hour period and work conditions do not allow for frequent breaks. This investigation also analysed responses to very light external loads (0.8kg, 1.6kg and 4kg) and the mass of the loads used may not be commonly found in industry. Therefore workers performing similar tasks may be taxed to a greater degree than demonstrated due to longer work duration and heavier loads manipulated. In addition, responses to only four load manipulation tasks were investigated where in industry many other tasks are performed.

CHAPTER II
REVIEW OF LITERATURE

INTRODUCTION

The maintenance of balance is an essential aspect of both static and dynamic task completion (Prioli *et al*., 2006) and is regulated through the input of sensory information (Mientjes and Frank, 1999; Radebold *et al*., 2000). To maintain balance, reactive and predictive strategies (Pollock *et al*., 2000) are used to achieve, maintain or restore the centre of gravity within the base of support (BOS) (Berg *et al*., 1989; King *et al*., 1994). Anticipated changes in equilibrium can, however, be opposed by anticipatory postural adjustments which occur prior to movement in the opposite direction to the expected perturbation (Bouisset and Zattara, 1990; Massion, 1992).

Spinal stability can be described as the ability of the spinal system to maintain an upright posture (Tanaka *et al*., 2009). The maintenance of spinal stability is important as a loss or decrease in spinal stability has been associated with the development of low back disorders (van Dieen *et al*., 2003c, Panjabi, 2003; Granata *et al*., 2004). The trunk muscles play a critical role in the maintenance of spinal stability (Cholewicki and McGill, 1996; McGill *et al*., 2003) through increased agonist and antagonist muscle contraction (Gardner-Morse *et al*., 1995; Cholewicki and McGill, 1996; Cholewicki *et al*., 1997; Gardner-Morse and Stokes, 1998). This may, however, be detrimental as increased co-contraction results in increased spinal loading (Granata and Marras, 1995a, Gardner-Morse and Stokes, 1998) and may also prevent motion (McGill *et al*., 2003).

Manual materials handling (MMH), especially lifting activities have been associated with the development of low back disorders (LBDs) and low back pain (LBP) (Dempsey, 1998; Lavender *et al*., 2006). Due to this, many ergonomic guidelines such as those created by Waters *et al*. (1993) have been developed but such guidelines do not take into account the underlying postural control mechanisms and muscle activity necessary for safe and efficient manual handling (Kollmitzer *et al*., 2002). In addition, it has been shown that postural control is task dependent (Horak and MacPherson, 1996) and therefore will change according to task demands. The relative contribution of the trunk muscles used to stabilise the spine during task completion are also affected by task constraints and change according to the direction and magnitude of trunk loading and will change throughout task completion (Cholewicki and van Vliet, 2002; McGill *et al*., 2003). Therefore it is necessary to investigate the muscles involved in postural control during a variety of tasks.

An increasing number of women are entering the work force and are performing tasks which have been traditionally been performed by men. This is particularly evident in South Africa as a result of the implementation of the Employment Equity Act of 1993. Differences in responses between males and females have not been established as research has tended to focus on male subjects (Lindbeck and Kjellberg, 2001). This research therefore included both male and female subjects in order to gain a representative sample. Anthropometric differences were however, accounted for during task completion.

BALANCE AND POSTURAL CONTROL

BALANCE

Balance is essential for the successful completion of motor tasks and is influenced by task difficulty. As such, it is necessary to study balance control in the context of task execution (Streepy and Angulo-Kinzler, 2002). Insight into the control of balance during task execution can be gained through measurements of the movement of the centre of pressure (COP) (Winter *et al*., 1996; Karlsson and Frykberg, 2000) which can be defined as the centre of the distribution of the total force applied to the supporting surface (Rougier, 2008). An increase in the movement of the COP can either indicate the movement of the centre of mass (COM) over the base of support (BOS) or an increase in the neuromuscular response necessary to maintain the position of the COP. In addition to this, increases in the amplitude of the COP movement during task execution can indicate the level of balance control as a movement is executed (Streepy and Angulo-Kinzler, 2002).

While standing, the body sways in all directions with variable muscular activity (Mello *et al*., 2007). This muscle activity is regulated by discrete neural stimuli and aims to maintain the projection of the (COM) within the limits of the BOS, thereby maintaining balance (Mello *et al*., 2007). Balance can be defined as the range of location of the

COP or the origin of the ground vector (Hof *et al*., 2005). In dynamic situations the velocity of the COM must be considered, as balance is possible in conditions where the COM is positioned outside of the BOS. For example, if the COM is outside the BOS but the velocity of the COM is directed towards the BOS, balance is possible. In the same way, if the COM is positioned above the BOS but the velocity of the COM is directed outward, balance may not be possible (Hof *et al*., 2005).

Postural control and balance involve the integration of information received from the visual, vestibular and somatosensory systems (Mientjes and Frank, 1999; Radebold *et al*., 2001.) as well as appropriately coordinated motor output at multiple joints which produce dynamic adjustments to a changing posture (Mergner and Rosemeier, 1998). The visual and vestibular systems provide the body with information regarding spatial orientation and the perception of motion. The somatosensory system gains information from muscle spindles, Golgi tendon organs and proprioceptors situated in joints. Further, mechanoreceptors detect pressure or shear forces produced by motion between the body and the supporting surface (Mergner and Rosemeier, 1998). This information is integrated and information regarding current body posture, spatial orientation and the position of the centre of gravity in relation to the BOS is fed back to a controller system (Massion, 1992; Silfies *et al*., 2003). If input from either of the above receptors is defective or reduced, body sway will increase and consequently muscle activity must increase in order to maintain equilibrium (Nardone *et al*., 1997).

Vision provides important information regarding the movement of the body within space and is therefore crucial in motor control and may override other sources of information (Schmidt and Lee, 1999). Central and peripheral vision contributes differently to the control of posture. Berencsi *et al*. (2005) propose that peripheral visual cues have a greater contribution in the control of quiet standing as it was found that body sway was reduced in conditions where peripheral visual cues were present in comparison to conditions where central visual cues were available. This proposal is consistent with earlier findings of Amblard and Carblanc (1980).

Static Balance

The initiation of joint stiffness through muscle activation is a simple mechanism through which the body's COM can be controlled during quiet standing. Winter *et al*. (1998) propose that in an upright posture, balance is regulated by a simple control mechanism which provides instantaneous corrections to balance and reduces the demands placed on the central nervous system. This proposition is referred to as the "stiffness theory" and suggests that the postural control system is passive and relies on muscular adjustments which are independent of sensory input (Winter *et al*., 1998; Mello *et al*., 2007). This model proposed by Winter *et al*. (1998) assumes that muscles act in a spring like manner which causes the COP to move synchronously with the COM as the body sways about a postural position. Winter *et al*. (1998) further suggest that in an upright posture, the body behaves as an inverted pendulum in which the body's COM is regulated through the movement of the COP under the feet. In the sagittal plane, stiffness control is present in the ankle plantar flexors and in the frontal plane in the hip abductors and adductors. Although theoretical (Morasso and Sanguineti, 2002) and experimental (Loram and Lakie, 2002) studies have confirmed that joint stiffness contributes to balance control, it is not the only mechanism responsible for the maintenance of equilibrium as anticipatory muscle actions initiated by the CNS also contribute to postural stability (Morasso and Sanguineti, 2002; Loram *et al*., 2005; Mello *et al*., 2007).

Dynamic Balance

The regulation of dynamic balance has received some attention but this has mainly occurred within the field of locomotion (Toussaint *et al*., 1997a; Zedka *et al*., 1998). In dynamic tasks two mechanisms are responsible for the control of movement and ensure successful task completion. The first mechanism is associated with the control of the ongoing movement while the second maintains equilibrium against both internal and external perturbations (Massion, 1992). Voluntary goal directed movement arises from internal forces produced by muscle contraction. Forces generated during movement also act on the body segments which support the movement and alter their position. The change in position of the body segments is expected and results in the active generation of anticipatory postural adjustments (APA) which aim to counteract the expected movement of the body segments

(Bouisset and Zattara, 1987; Lee *et al*., 1987). This type of balance control therefore preserves dynamic balance when the body is displaced along a trajectory towards a goal (Toussaint *et al*., 1997a) and will be discussed further at a later stage.

POSTURAL CONTROL

Postural control is a complex motor skill and can be defined as "the act of maintaining, achieving or restoring a state of balance during any posture or activity" (Pollock *et al*., 2000:404). The maintenance of an upright posture is a vital motor function (Deliagina *et al*., 2007) and postural control is essential for the maintenance of posture as well as task performance (Horak *et al*., 1997; Pollock *et al*., 2000). In humans, when the centre of gravity falls outside the base of support, the body has the ability to sense a threat to stability and in such instances, relevant muscles are activated to prevent falling (Horak, 1987; Pollock *et al*., 2000). Postural control strategies used are dependent on the assessment and control of many variables by the CNS. Therefore, the strategies used to maintain balance will differ according to an individual's goal, the environmental context and the task which is being performed (Horak and MacPherson, 1996; Horak *et al*., 1997). This view of postural control implies that balance control is a fundamental motor skill which is learnt by the CNS. This insinuates that postural control strategies can become more refined and efficient with training and practice (Horak and Nashner, 1986).

The control of balance has been associated with three broad classifications of human activity which include maintaining, achieving or restoring the centre of gravity within the base of support (Pollock *et al*., 2000). These classifications include the maintenance of specific postures such as sitting or standing, voluntary movements such as the movement which occurs between postures and the reactions to external disturbances such as a slip or fall (Berg *et al*., 1989; King *et al*., 1994).

Two main types of postural control strategies are used to restore or maintain balance. These are reactive strategies which are compensatory in nature and predictive strategies which are anticipatory in nature (Pollock *et al*., 2000). Combinations of both strategies may also be used. Predictive strategies occur in anticipation of predicted disturbances to balance and involve voluntary movement or increases in muscle activity. Reactive strategies on the other hand involve movement or muscle recruitment which is employed after an unpredicted disturbance to balance (Maki and McIlroy, 1997; Pollock *et al*., 2000). In addition to this, postural control strategies can be classified as being “fixed support" where the BOS is not altered during postural correction or “change in support” where the BOS is moved so that the centre of gravity intersects it (Pollock *et al*., 2000). The strategies mentioned above are depicted in Figure 1.

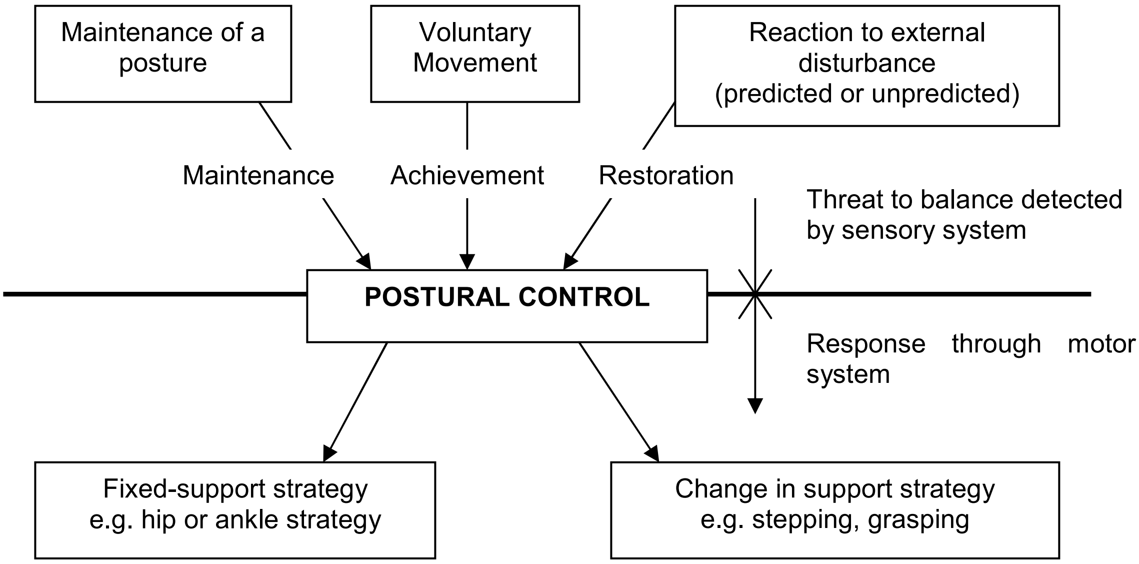


Figure 1: Postural control mechanisms (Pollock *et al*., 2000: 405).

Horak and Nashner (1986) identified distinct postural response sets termed the “ankle strategy” and the “hip strategy” which are used to maintain an upright posture. The ankle strategy makes use of distal to proximal muscle activation and is characterised by an inverted pendulum style of movement which incorporates motion of the knees and hips as well as the ankle. This strategy alters the body’s COM through torques generated primarily at the ankle and knee (Horak and Nashner, 1986; Horak, 1987; Runge *et al*., 1999). The hip strategy involves the use of early hip and trunk muscle activation and is characterised by large angular trunk accelerations. Further, this strategy adds torque generated at the hip to those of the ankle and knee (Horak and Nashner, 1986; Runge *et al*., 1999). The hip strategy is an important contributor to postural stability in the sagittal plane (Blackburn *et al*., 2003). In motion analysis studies, the hip and trunk are often viewed as one structure but postural control strategies in the frontal and sagittal plane result from motions arising from the hip and trunk (Horak and Nashner, 1986; Day *et al*., 1993; Rietdyk *et al*., 1999). Further, Blackburn *et al*. (2003) propose that trunk motion is as important as hip motion during postural control.

Swaying from the ankle or hip are commonly used and documented fixed support approaches (Pollock *et al*., 2000), while stepping or grasping with the hands are common change in support strategies (Pollock *et al*., 2000). It should be noted that the postural control strategies employed in each situation are dependent on both internal and external constraints. Internal constraints include the limitations of the biomechanical and nervous system, whereas external constraints are those imposed on the individual by the environment and the task (Horak, 1987). The postural control strategies discussed above can be seen in Figure 1.

Gahery (1987) classified postural control mechanisms during upright stance or limb movement into three elements namely postural preparations, postural accompaniments and postural corrections. The selection of the mechanism used is dependent on the need for the safe regulation of the body's COM and motor efficiency (Frank and Earl, 1990; Laessoe and Voigt, 2008). Postural reactions make use of sensory feedback and form the primary defence to unexpected external perturbations. Postural reactions are an efficient method of postural control but may not occur early enough to maintain posture. Postural preparations occur before movement and involve creating a more stable posture by increasing the base of support and stiffening joints through muscle co-contraction (Frank and Earl, 1990). This method of postural control creates a large safety margin but is considered an inefficient method of maintaining posture. The mechanisms associated with postural accompaniments occur concurrently or immediately before voluntary movement and involve anticipating the effect that the movement will have on posture. If task conditions are known, this method of postural correction can be safe and efficient (Frank and Earl, 1990). Research has focused on postural corrections in the form of reacting to different perturbations but little information is available regarding anticipatory strategies and postural preparations which occur in predictable perturbations (Commissaris and Toussaint, 1997; Laessoe and Voigt, 2008).

Anticipatory Postural Control

The human body can be considered as a multi-link structure and therefore any voluntary movement will cause equilibrium to be disturbed. This effect is amplified if movement occurs with added external loads such as when lifting objects (Kollmitzer *et al*., 2002). Anticipated changes in equilibrium, such as a shift in the position of the body's COM, are counteracted by anticipatory postural adjustments (APAs). The function of APAs is to generate joint torques which alters the centre of gravity in the opposite direction to the future perturbation (Bouisset and Zattara, 1990). APAs can be observed in postural muscles before postural perturbations and their assumed role is to counteract disturbance to equilibrium caused by movement in a feedforward manner (Bouisset and Zattara, 1990; Massion, 1992). APAs are involuntary and are incorporated into movement patterns to ensure that movement occurs in an accurate and smooth manner (Oddsson, 1990; Ioffe *et al*., 1996). When movement occurs in unstable conditions, the APAs implemented become more complex, involving a larger number of activated muscles and higher activation levels, as well as different muscle activation strategies such as co-contraction (Oddsson, 1990; Aruin *et al*., 1998).

Research has shown that the generation of APAs is dependent on a number of factors including the direction and magnitude of the perturbation (Lee *et al*., 1987; Aruin and Latash, 1996; Aruin *et al*., 2003), the stability of the body (Aruin *et al*., 1998; Nouillot *et al*., 2000), the characteristics of voluntary movements associated with the perturbation (Aruin and Latash, 1995), the ability to predict the future perturbation (Aruin *et al*., 2003) and the postural task (Aruin *et al*., 1998).

During quiet, stable standing, the body's equilibrium is not disturbed enough for this method of postural stabilisation to be used (Aruin *et al*., 1998). However, in conditions of postural instability, if initiated postural adjustments are stronger or greater than necessary, they may contribute to the perturbation itself. This may result in a perturbation in the opposite direction to the expected one which may outweigh the benefits of using strong APAs (Aruin *et al*., 1998). There have been conflicting results regarding the dependence of APAs on the stability demands of a postural task, as research has suggested that APAs associated with voluntary movements are extended or absent when posture is either highly unstable (Aruin *et al*., 1998; Slijper and Latash, 2000) or very stable (Nouillot *et al*., 1992). APAs may be absent when posture is stable as the CNS may consider the APA as an additional source of postural instability (Aruin *et al*., 1998). Therefore, according to the same authors, in conditions of postural instability the CNS may prevent the generation of APAs to ensure that the unstable equilibrium of the body is not exposed to additional potentially destabilising forces. This effect is more pronounced in conditions of greater levels of instability and if the direction of the instability and perturbation coincide (Aruin *et al*., 1998).

Santos and Aruin (2008), propose that in conditions of postural instability, the CNS may alter the form of APAs by using a number of strategies to maintain equilibrium. The first strategy involves increased activation of the dorsal and ventral muscles particularly those in the area of the perturbation. The second includes the co-activation of agonist-antagonist and lateral muscle pairs while the third strategy includes decreasing the strength of APAs in the trunk and leg muscles (Santos and Aruin, 2008). Research suggests that the muscle activation strategy selected by the CNS is dependent on the demands of the task and therefore, a strategy involving increased anticipatory activation of the trunk and leg muscles may be used in conditions of minor instability. However, anticipatory co-activation of muscles may be a preferred strategy in conditions of greater instability (Santos and Aruin, 2008).

The presence of APAs in dynamic tasks has been demonstrated by Toussaint *et al*. (1997a) and Commissaris and Toussaint (1997). The same authors propose that the APAs which occur during dynamic movements are similar to those which occur in static movements as they aim to counteract alterations in equilibrium during task completion (Commissaris and Toussaint, 1997; Toussaint *et al*., 1997b). A limited amount of research has focused on the development of APAs in tasks in which body segments have a dual function including task completion as well as postural control (e.g. Nashner and Forssberg, 1986; Hirchfield and Forssberg, 1991,1992). A whole body lifting task, would however allow for the examination of APAs in an ongoing multi-joint movement in which the legs serve both a focal and postural function (Commissaris and Toussaint, 1997).

Fatigue and postural control

Muscle fatigue affects postural control by increasing postural sway (Nardone *et al*., 1997; Vuillerme *et al*., 2002; Mello *et al*., 2007). Forestier *et al*. (2002) propose that during muscle fatigue, the associated reduction in postural control may be due to an inability of the ankle joint to stabilise the body and/or due to a decrease in the function of the proprioceptive system. When local muscle fatigue occurs, the effect of muscle action on the COP displacement is delayed which results in increased body sway (Mello *et al*., 2007). The observed delay may be due to the need for a more intense and longer period of neural stimulation after fatigue to produce effective muscle contraction (McArdle *et al*., 2001).

With regard to the trunk, Wilson *et al*. (2006) found that lumbar extensor fatigue resulted in changes in both reactive and proactive postural control strategies. Changes in proactive strategies included adopting a slight forward lean before the onset of the perturbation which resulted in the COP being displaced posteriorally (Wilson *et al*., 2006). Changes in reactive strategies included using a hip strategy to stabilise the body. Therefore in conditions of lumbar extensor muscle fatigue, postural control strategies shift towards using a greater degree of a hip strategy in order for equilibrium to be maintained (Wilson *et al*., 2006). Past research by Moore *et al*. 1988 does however suggest that the changes observed in reactive studies can be partly attributed to the forward lean adopted by the individual.

Vuillerme *et al*. (2007) propose that fatigue of the trunk extensor muscles induces a greater impairment to postural control in comparison to fatigue of the distal muscles such the ankle musculature. The same authors further suggest that trunk extensor muscle fatigue negatively affects the control of undisturbed stance especially along the anterior posterior axis and results in greater neuromuscular requirements to enable the control of standing accompanied by a reduction in postural performance (Vuillerme *et al*., 2007).

Due to the documented effects of fatigue on postural control, this investigation aimed not to induce fatigue during task completion. Thus, postural mechanisms were analysed in the most unaffected state possible.

SPINAL STABILITY

The term spinal stability can be used to describe the ability of the spine, paraspinal ligaments, core muscles and neuromuscular control system to maintain an upright posture (Tanaka *et al*., 2009). Spinal stability is essential in order to bear loads and allow movement, as well as to avoid pain and injury (Reeves *et al*., 2007). It has been shown that the risk of occupational low back disorders may be related to the musculoskeletal stability of the spine (Granata and Wilson, 2001; Anders *et al*., 2005) and therefore when balance is lost, the recovery strategy used must include the maintenance of spinal stability in order to avoid injury to the spine (Panjabi, 1992; Cholewicki and McGill, 1996; Cholewicki *et al*., 1997).

Spinal instability can be either the cause or the result of injury (McGill *et al*., 2003). Instability may occur due to a reduction in the mechanical integrity of load bearing tissue which would result in a decrease in spine stiffness. This would in turn cause a greater risk of unstable behaviour (McGill *et al*., 2003). Loss of spinal stability may cause excessive strain on the paraspinal tissues and this may be a potential cause of low back pain (van Dieen *et al*., 2003c; Panjabi, 2003; Granata *et al*., 2004). The stabilization system of the spine can be divided into three subsystems namely the passive, active and neuromuscular subsystems (Demoulin *et al*., 2007). The passive subsystem includes the vertebrae, intervertebral discs, ligaments, joint capsules and zygapophyseal joints. The active subsystem includes muscles and tendons which contribute to spinal stability. The neuromuscular system interprets information gained from mechanoreceptors located in active and passive structures (Panjabi, 1992; Demoulin *et al*., 2007).

Maintenance of spinal stability is achieved through feedback. In the spine system, three mechanisms are responsible for feedback control, namely the intrinsic properties of intervertebral joints, the intrinsic properties of the trunk muscles and the central nervous system (CNS) (Reeves *et al*., 2007). Feedback originating from intrinsic pathways is instantaneous, where feedback from the CNS, involving reflex and voluntary muscle actions involve delays which are associated with the time taken to sense a perturbation and to respond appropriately. Any combination of these control strategies may be used to stabilise the spine but the contribution of each mechanism changes depending on the size of the perturbation and the initial state of the spine system (Reeves *et al*., 2007). Tasks of an extended duration which require precise control, for example standing, tend to rely on reflex mechanisms to stabilise the spine as this method is metabolically efficient and also produces quicker responses to disturbances in equilibrium (Reeves *et al*., 2007). Due to the number of pathways which can be used to stabilise the spine, spinal stability can be maintained in a number of ways but it is suggested that each task has an optimal control strategy which reduces metabolic costs and/or enhances the performance of the system (Reeves *et al*., 2007). Under dynamic conditions, feedback control mechanisms aim to return the system's movement to its intended or original trajectory and in such cases when the system is perturbed, the forces generated by the mechanism of feedback control will be added to those already placed on the spine by the movement which is taking place. However, in a static condition for example, maintaining an upright posture, the forces acting on the spine system are balanced. Therefore if the system is perturbed, the force applied to the system to maintain stability will only include that from the feedback control mechanisms (Reeves *et al*., 2007).

Low back injuries are often caused by slips and falls which result in sudden loading or unloading of the lumbar spine (Troup *et al*., 1981; Omino and Hayashi, 1992). Therefore, lumbar spine stability is an important factor to consider when determining the response of the trunk to sudden loading (Gardner-Morse *et al*., 1995; Cholewicki and McGill, 1996). It has been hypothesised that injury caused by sudden loading or unloading may be a result of either inappropriate control responses of the trunk muscles (Panjabi, 1992; Magnusson *et al*., 1996; Radebold *et al*., 2000) or inadequate stabilization of the lumbar spine (Panjabi, 1992; Cholewicki and McGill, 1996). In most instances, muscle reflex responses are able to compensate for insufficient initial spinal stability. When carrying an external load, if spinal stability is sufficient prior to a sudden loading incident, only minimal adjustments by muscle reflexes may be necessary. If, however, the initial spinal stability is not great enough relative to the external load, a fast and strong reflex response may be vital in the prevention of intervertebral displacement and subsequent damage of soft tissues in sudden loading conditions (Cholewicki *et al*., 2000b).

In addition to the trunk muscles, spinal stability is influenced by factors including ligamentous support of the inter-vertebral joints, spine posture and the magnitude, direction and end-conditions of the external loads which are placed on the trunk (Cholewicki and McGill, 1996). Stiffness of the intervertebral joints impacts spinal stability as muscles which span these joints can provide additional stiffness when activated, as active muscle resists elongation (van Dieen *et al*., 2003a). Spinal instability may result from the inappropriate muscle during low level spinal loading (McGill *et al*., 2003). Therefore, stability or instability should be thought of as a state of the entire spinal system determined by the activation of all trunk muscles, passive joint properties, spine posture and loading conditions (Cholewicki and van Vliet, 2002).

Stability of the lumbar spine can be assessed using two different methods. The first relies on mathematical modelling of the spine and spine stability is quantified by estimating muscle forces generated during a task (Gardner-Morse *et al*., 1995; Cholewicki and McGill, 1996). The second method uses perturbation experiments in which trunk stiffness and stability are calculated from kinematic data acquired after a perturbation (Cholewicki *et al*., 2000a).

The role of the trunk musculature in spinal stability

As the spine is inherently unstable the trunk musculature plays a critical role in supporting the spine at all potential sources of instability (Crisco and Panjabi, 1991; Cholewicki and McGill, 1996; McGill *et al*., 2003). Each joint has a degree of inherent stability due to the stiffness of the ligaments surrounding the joint and the joint capsule, but the surrounding musculature greatly contributes to stability (Preuss and Fung, 2007). Stability provided by muscle is a result of muscle stiffness which is related to the biomechanical properties of the muscle as well as the level of muscle activation (Bergmark, 1989; Cholewicki and McGill, 1995). Therefore the surrounding musculature can greatly increase stability and when all stabilising muscles are recruited together, stability can be achieved even under high loading conditions (Gardner-Morse *et al*., 1995; Cholewicki and McGill, 1996). The prolonged muscle activity necessary to maintain the mechanical stability of the spine may, however, lead to the development low back pain (Panjabi, 1992; Cholewicki, 1993).

Traditionally, assessment methods used to quantify LBD risk have focused on spinal compression with compression forces below 3400N being considered safe for the majority of tasks (Waters *et al*., 1993). However, it has been demonstrated that injury to the spine due to spinal instability, may occur even with low levels of spinal loading (Cholewicki and McGill, 1996). Research has indicated that the critical load sustainable by the osteoligamentous structures of the lumbar spine without muscular support is approximately 90N (Crisco, 1989; Crisco and Panjabi, 1991; Crisco *et al*., 1992) but compressive loads in the range of 600N may be experienced during more demanding daily activities (McGill and Norman, 1985). Research has also highlighted the potential for spinal injury in conditions of low level loading (Crisco and Panjabi, 1991; Cholewicki and McGill, 1996). Cholewicki and McGill (1996) found that the stability of the lumbar spine increased during more demanding tasks (defined by compression forces imposed on the spine) and stability decreased during periods of low muscular activity. Therefore, the risk of injury due to instability increased with a decrease in muscular effort. This concept can be seen in Figure 2 and would aid in the explanation of how injury occurs at low levels of loading and further demonstrates the importance of the spinal musculature in the prevention of injury by increasing spinal stability. Considering this, mechanical stability of the spine should be considered at all levels of spinal loading (Cholewicki and McGill, 1996).

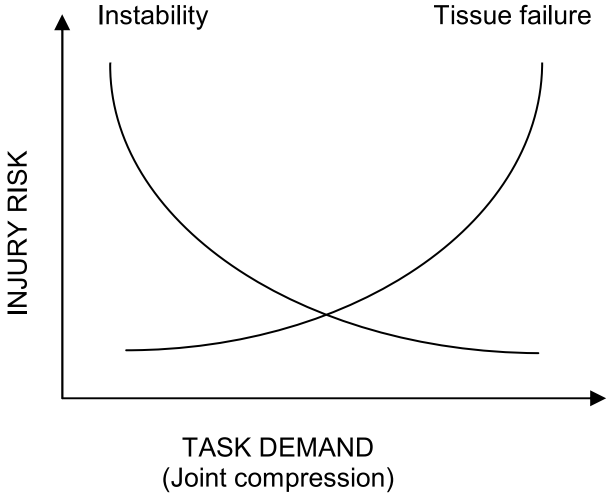


Figure 2: Injury risk to the spine due to tissue failure and spinal instability (Cholewicki and McGill, 1996: 8).

Analytical analyses have shown that spine stiffness alone is insufficient to stabilise the trunk and that activated muscle stiffness is necessary to maintain trunk stability (Bergmark, 1989; Crisco and Panjabi, 1991; Gardner-Morse *et al*., 1995; Gardner-Morse and Stokes, 1998). Spinal stability is actively controlled through the regulation of force in the surrounding muscles (Cholewicki *et al*., 2000b). Muscle force is approximately linearly proportional to muscle stiffness due to the increased number of activated cross-bridges during muscle activation (Crisco and Panjabi, 1991; Cholewicki and McGill, 1995). Cholewicki and van Vliet (2002) propose all of the trunk muscles contribute to spinal stability and that it is not possible to identify one muscle that is the most important in maintaining the stability of the lumbar spine, as the relative contribution of each muscle will depend on the direction and magnitude of trunk loading. Therefore, the contribution of muscles to spinal stability will change under different loading conditions, with the recruitment patterns of other trunk muscles and will also change throughout task completion (Cholewicki and van Vliet, 2002; McGill *et al*., 2003).

Although co-activation of agonistic and antagonistic trunk muscles increase spinal stiffness and stability (Gardner-Morse *et al*., 1995; Cholewicki and McGill, 1996; Cholewicki *et al*., 1997 Gardner-Morse and Strokes, 1998), muscle co-activation results in greater compression forces which tend to destabilise the spine (Granata and Marras, 1995a; Gardner-Morse and Stokes, 1998). Excessive stiffness and muscle co-activation will result in substantial increases in spinal loading and may also prevent motion (McGill *et al*., 2003). “Sufficient stability” is therefore a concept which involves necessary muscle stiffness produced by muscle contraction in order to ensure stability, together with a degree of extra stability to form a safety margin for injury prevention (McGill *et al*., 2003). Large muscular forces are rarely needed to maintain sufficient stability as modest force generation results in a rapid increase in joint stiffness which aids in stabilization (McGill *et al*., 2003). Sufficient stability of the lumbar spine in un-deviated postures is generally achieved through relatively small levels of co-activation of the paraspinal and abdominal wall muscles and consequently, sufficient stability can be maintained in all activities with continuous low levels of muscle activation (Cholewicki and McGill, 1996; Cholewicki *et al*., 1997). Therefore when performing daily living activities the safety margin is not compromised due to insufficient strength but may be exceeded due to decreases in endurance capacity and inadequate control of movement (McGill *et al*., 2003). The level of sufficient stability needed to prevent injury continuously changes as a function two variables. Firstly the three-dimensional torques necessary to maintain posture and secondly the stiffness needed to anticipate unexpected loads; to prepare for the need to move quickly; or to ensure stiffness of a joint which may be compromised due to injury (McGill *et al*., 2003).

Sustained muscle contraction results in a decrease in the force generating capacity of muscle (McArdle *et al*., 2001) and therefore, in order to maintain a desired level of force, muscle activity increases which in time may lead to fatigue (Petrofsy *et al*., 1982; Duchene and Goubel, 1990). Trunk muscle fatigue has been shown to decrease muscle coordination (Potvin and O'Brian, 1998) resulting in greater spinal instability and the production of movement patterns that are not usually observed in conditions of rest (Mital *et al*., 1993; Grondin and Potvin, 2008).

Posture and spinal stability

It has been established that body posture during manual handling tasks impacts spinal loading (Gallagher *et al*., 2002). Previous studies have shown that trunk muscle activation differs significantly when tasks are performed in restricted postures (Gallagher *et al*., 1988, Gallagher and Unger, 1990, Gallagher *et al*., 1994) and trunk recruitment patterns have shown to differ when loads are increased (Fathallah *et al*., 1998; de Looze *et al*., 1999; Davis and Marras, 2000). It is however, not clear whether the associated changes in muscle activity in these two conditions are independent or whether they interact (Gallagher *et al*., 2002).

In conditions with high levels of spinal loading, appropriate muscle recruitment provides sufficient spinal support (Punnett *et al*., 1991; Granata and Marras, 1999) but changes in posture may limit the ability of the neuromuscular system to maintain spinal stability (Granata and Wilson, 2001). Further, Granata and Wilson (2001) propose that postural risk factors associated with LBDs may be partially explained by spinal stability and therefore, in lifting situations it is necessary to estimate spinal stability as a function of lifting posture.

Trunk posture during lifting has been identified as a risk factor for the development of LBDs (Marras *et al*., 1995). In addition, retrospective studies have concluded that lifting, accompanied by twisting (Kelsey *et al*., 1984), lateral bending (Punnett *et al*., 1991) and asymmetric postures (Marras *et al*., 1993) increase the risk of LBD. When performing tasks in high risk postures, such as those mentioned above, research has indicated that antagonistic muscles are recruited in order to maintain spinal stability (Cholewicki *et al*., 1998).

Maintaining fully flexed postures for extended durations has also been linked to the development of LBDs (Shin and Mirka, 2007). The degree of lumbar flexion is controlled to a certain extent by the erector spinae muscles but at approximately 60° of lumbar flexion, (Burgess-Limerick, 2003) the erector spinae muscles become myoelectrically silent which is referred to as the flexion-relaxation phenomenon (Kippers and Parker, 1984; Christopher *et al*., 2005). In a fully flexed posture, body weight is supported by the intervertebral discs together with the extension moment generated by the spinal ligaments and the extensor muscle- tendinous units (McGill and Kippers, 1994). Maintaining flexed postures for a prolonged period may result in the deformation of the spinal tissues, which can cause a reduced resistance to a forward flexion moment (Adams and Dolan, 1996; Parkinson *et al*., 2004; Solomonow, 2004). In addition, prolonged flexion results in the inability of the spinal ligaments to return to resting length when unloading occurs. This may result in a temporary loss of spinal stability which may lead to an increased risk of injury during spinal loading in any posture (McGill, 1997). Burgess-Limerick (2003) further propose that lifting from postures involving extreme lumbar flexion can potentially cause damage to the spinal ligaments and intervertebral discs especially when lifts are combined with lateral flexion and rotational movements.

Postural control of the trunk can be influenced by the magnitude and direction forces exerted by the trunk. Increased force production during flexion and extension movement results in increased co-activation of the trunk flexor and extensor muscles (Granata *et al*., 2005; Lee *et al*., 2007). Although co-contraction results in increased spinal stability, it may have a negative effect on trunk postural control (Lee *et al*., 2008) as increased co-activation causes greater variability in force output (Christou *et al*., 2002; Hamilton *et al*., 2004). Trunk flexion and extension exertions may also influence postural control of the trunk and altered trunk postural control could potentially contribute to the development of low back disorders which occur during these types of movements (Radebold *et al*., 2001; Lee *et al*., 2008). Investigations regarding the effects of trunk exertions on trunk postural control are however limited (Lee *et al*., 2008) but this concept has been investigated during unstable sitting where it was found that increased trunk exertion caused a decrease in trunk postural control. Trunk flexion exertions were also associated with poorer levels of postural control when compared to extension exertions (Lee *et al*., 2008).

Influence of low back pain on spinal stability

Low back pain (LBP) is often caused by injury to the spinal ligaments or intervertebral discs which result as a consequence of mechanical overloading (Pope and Novotny, 1993; Adams and Dolan, 1997; van Dieen *et al*., 1999) and these injuries have a great impact on the mechanical behaviour of the spine (van Dieen *et al*., 2003b). Research has shown that individuals with LBP have reduced trunk muscle force (Lee *et al*., 1995; Takemassa *et al*., 1995) and reduced endurance capability of the trunk extensor muscles in comparison to healthy individuals (Biedermann *et al*., 1991; Luoto *et al*., 1995). Based on these findings, LBP patients have a reduced ability to rapidly generate muscle force which results in a limited ability to correct trunk perturbations and prevent spinal instability (van Dieen *et al*., 2003b). Further, it has been proposed that LBP patients require additional muscular stabilization of the spine and this may be necessary for a number of reasons. Firstly, LBP patients may have reduced passive stiffness of the spine as a consequence of damaged discs or ligaments. Secondly, the ability to respond to and correct balance perturbations may be reduced due to decreased muscle force and lastly because sensorimotor integration may be altered in LBP patients which slows down corrective responses (van Dieen *et al*., 2003b).

Spinal ligaments have an important sensory role by providing feedback on joint position and, therefore, injury to spinal ligaments is likely to result in a disturbance of trunk equilibrium (Sjolander *et al*., 2002). Decreased passive stiffness of the spine associated with LBP increases the risk of spine buckling during balance perturbations. In addition, changes in the muscular and control system would decrease the ability of LBP patients to respond appropriately to trunk perturbations.

This has been confirmed in the literature as it has been shown that LBP patients respond less effectively to trunk perturbations (Magnusson *et al*., 1996; Radebold *et al*., 2000). Therefore alterations in recruitment patterns in LBP patients may be functional, resulting in increased levels of trunk stiffness. This in turn would decrease the number of perturbations which may cause injury in LBP patients (van Dieen *et al*., 2003b).

MANUAL MATERIALS HANDLING AND LOW BACK DISORDERS

MANUAL MATERIALS HANDLING

Manual materials handling (MMH) tasks include the lifting, lowering, pushing, pulling and carrying of materials (Ciriello *et al*., 1999). MMH, especially lifting, has been closely associated to the development of low back injuries and low back disorders (Norman *et al*., 1998; Dempsey, 1998; Marras *et al*., 2000b; Lavender *et al*., 2006) but despite increases in mechanisation MMH continues to be an essential component of work (Davis and Marras, 2000). MMH tasks impose stresses on the worker which manifest as strain on the muscular and cardiovascular systems (Dempsey, 1998). If this strain is greater than the tolerance of either system, fatigue, discomfort and injury may occur. Therefore task demands and worker capabilities should be balanced such that performance can be optimised and the risk of injury minimised as far as possible (Dempsey, 1998).

In order to control the risks associated with MMH activities, researchers from various fields have developed work criteria which should not be exceeded during work. The proposed criteria are based on the principles of biomechanics, physiology and psychophysics. Epidemiological criteria have also been developed but are often used to evaluate proposed criteria and investigate the potential causes of injury associated with MMH activities (Dempsey, 1998). The biomechanical approach aims to design tasks which are within the limits of the musculoskeletal system. The two most commonly used criteria in this approach are compression forces at the level of L_5/S_1 and maximum joint torques. Tasks are assessed according to these criteria through either a static or dynamic biomechanical model which provides estimates of either compression or joint torque (Dempsey, 1998). The physiological approach focuses on the physiological responses to different tasks. Commonly criteria in this approach is the amount of energy expended during a task, expressed in either kcal/min or oxygen consumption as a percentage of maximum aerobic capacity. The psychophysical approach aims to design tasks which are deemed “acceptable” to the majority of workers performing a task (Dempsey, 1998). At this point it should be noted that each of the above approaches have limitations and, by their nature, the criteria proposed are conflicting and as such it may not be possible to satisfy all the proposed criteria simultaneously (Dempsey, 1998).

Although MMH, especially lifting, has been widely investigated and lifting guidelines have been developed which aim to reduce the risks associated with MMH, the guidelines do not take into account the postural control mechanisms associated with increased load and body movement which are necessary for safe and efficient manual handling (Kollmitzer *et al*., 2002). The mechanisms of postural control employed during lifting tasks place an individual at an increased risk for the development of LBDs as the stress of the task is added to the stress which is already placed on the body by the muscle activity necessary to maintain equilibrium (Reeves *et al*., 2007). Even though this has been established, only a few studies have investigated the postural control strategies adopted while performing manual handling tasks (Toussaint *et al*., 1997a, b; Toussaint *et al*., 1998). In addition to this, ergonomic analyses assume that repeated exertions will result in the same amount of strain when workplace factors remain constant. Analyses have, however, shown a great potential for variability during repeated exertions and this variability may affect the interpretation of ergonomic assessment results (Granata *et al*., 1999).

LOW BACK DISORDERS

Low back disorders (LBDs) have been identified as the most common and expensive work related musculoskeletal disorder (Dempsey, 1998) and it has been proposed that up to 80% of individuals will experience low back pain (LBP) at some time in their lives (Plante *et al*., 1997). In order to reduce the occurrence of LBP and LBDs it is necessary to understand the mechanisms of injury associated with the spine. Biomechanical logic is a means through which loads placed on the spine can be quantified and is based on the load-tolerance relationship (Marras, 2000).

a)

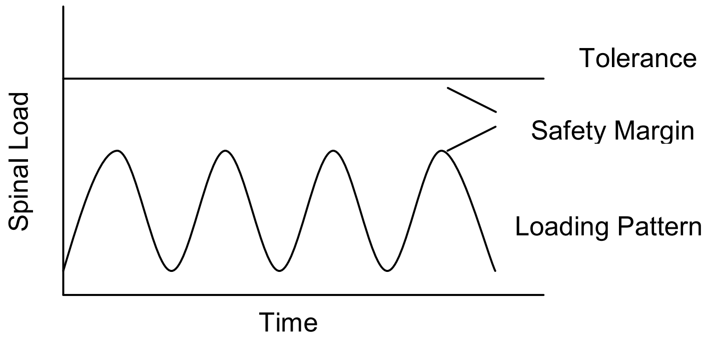


Figure 3 (a): Biomechanical logic (Marras, 2000: 887).

The load tolerance relationship assumes that a quantifiable load is placed on the spine and if the load is below the tolerance limit of the tissue, injury is not expected to occur (Marras, 2000). From this information, a safety margin can also be quantified which can be described as the difference between the imposed load and the tolerance of the tissues (McGill, 1997). This concept is illustrated in Figure 3 (a).

Spinal injury is a result of mechanical overloading and is determined by the characteristics of the external load and the properties of the tissue on which loading occurs (McGill, 1997). Injury to the spine occurs when a load applied is greater than the strength of the tissue (McGill, 1997). A once-off applied load can result in injury but it has been proposed that only few low back injuries occur in this manner (McGill, 1997; Marras, 2000). LBD are cumulative in nature and, therefore, it is suggested that LBDs are caused by increases in spinal loading which occurs throughout the day. This is accompanied by a decrease in tolerance and, when loading becomes greater than the tolerance of the soft tissues, injury occurs (McGill, 1997). Consequently, spinal loading at the beginning of a work day may be below tolerance limits but will increase throughout the work shift (Marras *et al*., 2006). Injury may also occur during repetitive tasks as according to Mirka and Marras (1993) repetitive tasks may result in variable loading which may result in loads greater than the tolerance being applied to spinal tissues. Although spinal loading may result in injury, optimal loading patterns can result in a training effect which can cause an increase in tolerance (Marras, 2000). The mechanisms of injury described above can be seen in Figure 3 (b) and (c).

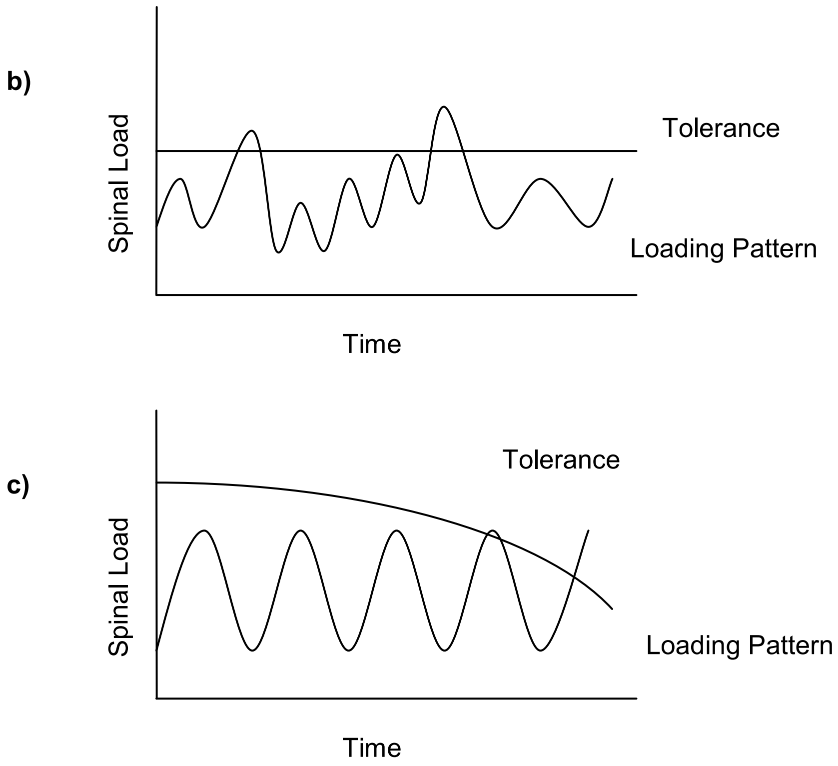


Figure 3: Cumulative trauma due to variability in loading b) and cumulative trauma due to tolerance decrease c). (Marras, 2000: 887).

Personal risk factors for LBDs

Personal as well as occupational factors may place individuals at risk for the development of LBDs. Personal risk factors for LBDs include age, gender, anthropometry and smoking. LBD usually affects relatively young individuals with the highest frequency of symptoms occurring between the ages of 30-35 and absenteeism from work due to back pain increases with increasing age (Andersson, 1997). It has also been suggested that for males, LBD risk peaks at approximately 40 years of age where the risk for women peaks between 50 and 60 (Marras, 2000). Although anthropometry has been identified as a personal risk factor, there is little agreement in the literature as sitting height (Ferguson and Marras, 1997) and obesity (Deyo and Bass, 1989) have been suggested as potential risk factors associated with low back disorders. Therefore, personal risk factors for the development of LBD may exist but the strength of these correlations are low and in most cases are beyond the control of individuals and society (Marras, 2000). It is, however, important to note that such factor may interact with occupational risk factors and contribute to overall risk (Marras, 2000).

Task related risk factors for LBDs

Traditionally, work place risk factors have focused on heavy physical work, static working postures, bending and twisting, whole body vibration and lifting and forceful movements (Marras, 2000). Critical reviews performed by Bernard (1997) and Hoogendoorn *et al*. (1999) found strong evidence of risk for LBD with lifting and forceful movements, bending and twisting as well as whole body vibration where moderate evidence of risk was found with regard to heavy physical work. Psychosocial factors such as job dissatisfaction, monotony of work, limited job control and lack of social support may also contribute to LBD risk (Marras, 2000) but although these psychosocial factors have been recognised as contributing to LBD risk, no studies have investigated the physical and psychosocial risk factors simultaneously (Davis and Heaney, 2000).

Awkward working postures

The posture adopted during task completion affects a workers ability to reach, hold and use equipment and also impacts how long an individual is able to perform a task without experiencing adverse effects such as fatigue, discomfort and musculoskeletal disorders (Chung *et al*., 2003). The association between awkward working postures and the symptoms of musculoskeletal disorders has been established by numerous studies but this research has tended to focus on postures of the upper body including those of the wrist, shoulder, neck and lower back (van Wely, 1970; Putz-Anderson, 1988; Armstrong *et al*., 1993). Although the association between leg postures and musculoskeletal disorders has been documented (van Wely 1970; Gallagher *et al*., 1988) leg postures have received little attention in research. Research has however shown that leg postures have a strong association with postural stability and mobility (Kirby *et al*., 1987) and are also important in the evaluation of postural stress (Chung *et al*., 2003).

Although the human body is highly adaptable, working in awkward postures or in conditions with high environmental demands, the musculoskeletal system may experience substantial performance limitations (Gallagher, 2005). The posture adopted during manual handling tasks has a significant impact on spinal loading

(Gallagher *et al*., 2002) and studies have confirmed that restricted postures result in significant changes in trunk muscle activity (Gallagher *et al*., 1994; Gallagher *et al*., 1990; Granata and Wilson, 2001) which may impact spinal loading in different postures.

Trunk motion

Trunk motion as well as three dimensional trunk velocity and acceleration have been identified as potential risk factors in the development of LBDs (Marras *et al*., 1993; Norman *et al*., 1998; Marras *et al*., 1995). Dynamic motion plays a dominant role in the development of LBDs especially when movement occurs in multiple planes simultaneously. However, the mechanism of injury associated with dynamic trunk motion is not well understood (Davis and Marras, 2000). Trunk strength and spine structural loading are two major categories of risk for LBDs. LBDs resulting from a lack of strength would exceed the muscular capabilities of the spine while those resulting from structural loading are caused by forces placed on the spinal structures which are greater than the tolerance limits of the associated structures. Although it is known that both these factors can cause LBDs it is thought that each is associated with different mechanisms of LBD (Davis and Marras, 2000). Factors contributing to structural loading include the biomechanical factors which contribute to spinal loading such as intra-abdominal pressure, muscle activity, the moment imposed as well as the external loads placed on the spine due to load carriage (Davis and Marras, 2000). The external load placed on the spine is counteracted by the trunk muscles (internal load) (McGill and Norman, 1986; Marras and Granata, 1997) and therefore, increased trunk moment is accompanied by an increase in muscle activity which would result in greater compression and shear forces placed on the spine (Davis and Marras, 2000).

When performing dynamic trunk movements, individuals adopt a motion profile and muscle co-activation patterns based on previous experience (Marras *et al*., 2000b). It has been demonstrated by numerous investigations (Granata and Marras, 1995a, b; Gallagher, 1997; Dempsey *et al*., 1998) that on average, an individual’s strength is reduced by 10-30% when exertions are performed dynamically. Therefore, at the same exertion level, a dynamic movement would be closer to the muscular tolerance limit compared to a static exertion. Consequently, dynamic movements are associated with a greater risk of muscular injury (Davis and Marras, 2000). With regard to muscle activity, both agonist and antagonistic muscle activity has been shown to increase with increased dynamic trunk motion (Marras *et al*., 1984; Cresswell and Thorstensson, 1989; Granata and Marras 1995b; Gallagher, 1997) which indicates a greater overall activation of the trunk musculature. Greater levels of co-activation may increase spinal loading as antagonistic contraction would add to spinal loading without contributing to counteracting the external moment generated by an external load (Davis and Marras, 2000). Therefore, dynamic motion contributes significantly to the compressive forces placed on the spine and this appears to be independent of the type of motion. In addition, the decrease in strength, functional capacity and increase in spinal loading associated with dynamic trunk movement may be a result of muscle co-activation (Davis and Marras, 2000).

Trunk dynamics are further influenced by factors such as sex, external load mass, task asymmetry, coupling and the way in which a task is performed. Little is known about the sex differences associated with trunk motion as most research regarding these differences has been confined to strength assessments (Davis and Marras, 2000). Increases in external load mass have been found to decrease trunk motion in the sagittal plane (Gagnon and Smyth, 1991; Ferguson *et al*., 1992; Granata *et al*., 1999; Lin *et al*., 1999). Task asymmetry also affects trunk dynamics as asymmetric tasks have been found to have greater three-dimensional trunk velocities (Ferguson *et al*., 1992; Allread *et al*., 1996; Maras and Davis, 1998). Fatigue is also an important aspect to consider as fatigue has been shown to influence trunk motion, muscle activity and associated spinal loading. Further, it has been proposed that fatigue causes a decrease in trunk motion in the sagittal plane and an increase in trunk motion in the lateral and twisting planes (Parnianpour *et al*., 1988; Gomez *et al*., 1991).

LOAD MANIPULATION

During load manipulation the total force acting on the spine is comprised of a combination of both external and internal forces. External forces generate moments about the spine caused by an external load, body segment weight and lever length

(Ferguson *et al*., 1992). Internal forces, generated by muscles, intra-abdominal pressure and passive components of the body act to counteract the effect of external forces placed on the body. Due to the body's mechanical disadvantage, internal forces are much greater than external forces (Ferguson *et al*., 1992). During lifting, the activation of various combinations of muscles may be used to generate a trunk moment and, therefore, in order to optimise muscle recruitment, a unique muscle activation pattern may be determined (Granata and Marras, 1995b).

Lifting

Lifting is one of the most common manual handling activities and accounts for up to 40% of manual handling tasks (Ciriello *et al*., 1999). Due to the cumulative nature of LBDs, repetitive lifting has been identified a risk factor for the development of LBDs (Frymoyer *et al*., 1983; Marras *et al*., 1993; Marras *et al*., 1995; Marras, 2003). Trunk flexion results in a high bending moment at the osteoligamentous lumbar spine (Adams and Dolan, 1991) and the act of lifting simultaneously generates high compressive forces on the spinal structures. Lumbar flexion may be further increased under a number of conditions such as when objects are lifted with straight legs, when the object is bulky or if the object is placed far away from the body (Dolan *et al*., 1994). The erector spinae muscles play an important role in protecting the spine from excessive flexion but if these muscles become fatigued as may occur during repetitive lifting; these muscles would be less able rapidly to generate force which may increase the lumbar bending moment (Dolan and Adams, 1998). Repetitive stretching of the lumbar viscoelastic tissue which occurs during cyclic flexion-extension movements significantly alters the mechanical and neuromuscular functions of the lumbar structures which in time may lead to the development of LBDs (Olson *et al*., 2009). In addition, repetitive loading can reduce the tolerance of the spinal tissues to a point at which previously acceptable loads could result in injury (Potvin, 2008).

During lifting tasks, the trunk and hip muscles provide the extensor torque necessary to return the upper body to an upright position. The trunk muscles control this movement while the hip muscles provide mechanical advantage during lifting (Abdoli-E and Stevenson, 2008). Lifting a load places the body at a mechanical disadvantage as the distance of the trunk extensor muscles is smaller than the distance from the external load to the spine. Therefore, in order to counteract an external load, the extensor muscles must generate greater forces than the weight of the load in which case internal loading must be greater than external loading (Jorgensen *et al*., 2003). During submaximal lifting, the neuromuscular system does not necessarily recruit those muscles with greatest mechanical advantage (Zetterberg *et al*., 1987) and this may influence spinal loading (Granata and Marras, 1995b).

During lifting, the erector spinae muscles are aided by other mechanisms in order to provide a sufficient extensor moment to counteract the external load including inter-abdominal pressure (van Dieen *et al.,* 2003a); muscle ligaments and discs (McGill and Norman, 1986); the additional recruitment of agonistic muscles (Granata and Marras, 1995b; Davis and Marras, 2000) and the hydraulic effect of the cylindrical trunk (Bogduk, 1999). The combination of these mechanisms has been shown to contribute up to 30% of the required extensor torque (Abdoli-E and Stevenson, 2008).

Lifting with an extended horizontal reach is a common activity in industry (Ciriello, 2007) and increased horizontal distance results in a decrease in maximum acceptable weights lifted for both males and females (Ciriello *et al*., 1993; Ciriello, 2003; Ciriello, 2007). Increased horizontal reach places the body at a greater mechanical disadvantage due to the increased moment arm of the external load thereby increasing the external force placed on the body (Ferguson *et al*., 1992; Jorgensen *et al*., 2003). As a result, the torso extensor muscles increase force production to counteract the increased external force (Jorgensen *et al*., 2003) and the increase in internal forces of the ultimately leads to increased spinal loading (Ferguson *et al*., 1992).

Evidence has shown that the variability in spinal load and biomechanical performance associated with repetitive lifting may influence the risk of low back disorders (Granata *et al*., 1999). Mirka and Marras (1993) demonstrated that biomechanical variability influences the number of repeated exertions which may exceed lifting guideline limits.

Asymmetrical lifting

Axial twisting of the torso has been identified in a number of epidemiological studies as a significant risk factor for occupationally related LBDs (Kelsey *et al*., 1984; Punnett *et al*., 1991; Marras *et al*., 1993). LBD risk increases when twisting velocities of even a low magnitude are present during a task (Marras *et al*., 1993). Additionally, asymmetrical lifting has been associated with a decrease in trunk strength (Marras and Mirka, 1989; Ferguson *et al*., 1992) a reduction in maximum acceptable weights of lift (Mital and Fard, 1986; Mital *et al*., 1989) and the development of more complex trunk motion patterns (Ferguson and Marras, 1992; Ferguson *et al*., 1992; Allread *et al*., 1996).

During asymmetrical lifting, external loads are supported by the oblique muscles rather than the erector spinae muscles (Marras and Mirka, 1992). Asymmetric lifting tasks also result in increased antagonistic trunk muscle co-activation (Marras and Mirka, 1990; Marras and Mirka, 1992) which causes greater spinal compression when lifts are performed in this manner (Marras and Sommerich, 1991; Granata and Marras, 1993). During asymmetric lifting, altered muscle recruitment is necessary in order to counteract the lateral moments generated by asymmetric trunk motion as well as to maintain body equilibrium (Schultz *et al*., 1982a, b). Thus, at increased exertion levels or in dynamic conditions, the stability requirements provided by antagonistic co-contraction may cause spinal compression values which are closer to injury tolerance levels (Granata and Wilson, 2001). As such, lifting capacity decreases and spinal loading increases when lifts are performed asymmetrically (Marras and Davis, 1998).

Due to the increased need to control and stabilise the trunk during asymmetrical tasks, increased levels of asymmetry have been found to increase levels of muscle coactivity (Marras and Mirka, 1992; Granata and Marras, 1995a; Cholewicki and McGill, 1996). Additionally, task asymmetry has been associated with variable spinal loading, increasing the number of exertions which may exceed biomechanical tolerance levels resulting in a greater risk of LBD development during asymmetrical tasks (Granata *et al*., 1999).

Although it has been presumed that lifts occurring on the right and left of the sagittal plane result in the same degree of spinal loading and trunk muscle activity, biomechanical literature suggests that spinal loading may be influenced by the direction of the lift asymmetry (Marras and Davis, 1998).

Lowering

Research suggests that the body responds differently to lowering tasks in comparison to lifting tasks. Perceived exertion and back extensor muscle activity is lower during lowering tasks in comparison to lifting tasks (Henriksson *et al*., 1972; Kumar and Davis, 1983; Marras and Mirka, 1989; de Looze *et al*., 1993; Cresswell and Thorstensson, 1994). Additionally, it has been demonstrated that individuals are stronger (by up to 50%) but have lower muscle activation levels during lowering motions (Davis *et al*., 1998) This indicates that the ability of the trunk muscles to produce force differs between the two conditions. Spinal loading is also affected by the type of motion analysed as Davis *et al*. (1998) found smaller anterior-posterior shear forces and greater compression forces during lowering exertions when compared to lifting exertions.

Postural control during lifting

Bi-manual whole body lifting is a task which involves both the execution of a flexion moment as well as the necessity to control dynamic balance within gravity (Toussaint *et al*., 1997a, b). In whole body lifting tasks, the maintenance of equilibrium is challenged due to the pick up of an object positioned in front of the body. The addition of an extra load to the body results in a forward shift of the COM in relation to the base of support (Commissaris and Toussaint, 1997; Toussaint *et al*., 1997b). In addition, the inertia of the lifted object tends to decelerate the extension of the body towards the final upright posture. This does not result in a change in the position of the COM in relation to the BOS but could disrupt the smooth extending motion of the body if these effects are not adequately anticipated (Commissaris and Toussaint, 1997). Anticipatory postural adjustments can, however, be used to counteract the above effects prior to their occurrence (Commissaris and Toussaint, 1997). Therefore, for a smooth lift to occur, adequate anticipatory control, which is appropriate for the mass of the load which is lifted, is necessary. As the mass of the load can only be accurately determined once the load has been picked up, the motor program parameters must be selected before the lift is performed. As a result, prospective control processes are adopted using memory of relevant object characteristics in order to scale responses appropriately (Ghez *et al*., 1991; Johansson and Cole, 1992; Gordon *et al*., 1993; Toussaint *et al*., 1997b).

When the primary motor task involves maintaining a static position or moving in a cyclic pattern, increasing stability may require extra muscular effort in order to reduce variability around a position or pattern and to respond quickly to perturbations (Houdijk *et al*., 2009). In biomechanical analyses, movements are typically analysed by averaging data over a number of repetitions. This method of analysis only provides information regarding the underlying average movement pattern and does not account for the muscular effort related to movement variability and co-contraction. In addition, this method does not reveal the extra muscular effort associated with balance control (Houdijk *et al*., 2009).

MEASUREMENT OF POSTURAL CONTROL

Posturography refers to the description of posture and most frequently, posturography techniques are used to investigate active and passive regulation of balance under a variety of conditions (Visser *et al*., 2008). Posturography can either be static or dynamic in nature. Static posturography involves the assessment of postural control in a relatively unperturbed state (usually in quiet stance) where dynamic posturography makes use of experimentally induced perturbations to balance (Furman *et al*., 1993).

Posturography allows for a detailed measure of postural responses by using a variety of quantifiable measures of postural control which include kinetic measures (causes of movement), kinematic measures (actual movement of body segments) or measures of muscle activity, for example electromyography. A combination of the above methods is preferable as complimentary information is provided by these measures (Visser *et al*., 2008).

Force platform

Postural control or postural sway is typically quantified through the use of a force platform. Posturography has been widely used to identify impairments in balance and postural control in the healthy and in those with a variety of neuromuscular disorders (Gregoric *et al.*, 1981, Alder *et al.*, 1986, Byl and Gray, 1993). In such studies, the movement of the centre of pressure, also referred to as body sway, was recorded during quiet stance. More recently, this method has shown that patients with low back pain have poorer postural control compared to healthy controls (Byl and Sinnott 1991, Takala *et al.*, 1997, Mientjies and Frank, 1999). The movement of the centre of pressure can be used to determine summary statistics such as the total distance travelled per second. However, these measures do not reflect physiological properties of the postural control system (Silfies *et al.*, 2003). Body sway, which occurs in the horizontal plane, contains valid information regarding postural stability (Karlsson and Frykberg, 2000). Impairments in the balance system may reduce stability resulting in an increase in body sway and/or an altered movement strategy (Horak and Nashner, 1986).

Electromyography

Until it can be verified which muscles are used in different motor tasks, deductions about their function is made from the anatomical arrangement of the muscle. Electromyography (EMG) is an invaluable tool which enables one to verify the involvement of muscles during task completion objectively (Basmajian and de Luca, 1985; Andersson *et al.*, 1996). EMG is a technique which records the changes in the electrical potential of a muscle during muscle contraction (Gleeson, 2001) and is a valuable tool which can be used when investigating function studies of movement co-ordination (Goebel, 2005). Changes in postural responses can be identified through the use of EMG by measuring onset latencies, response amplitudes or postural strategies (Visser *et al.*, 2008).

The EMG signal can be measured using two techniques including surface and needle electrodes. The use of surface electrodes is a non-invasive method of detecting superficial muscle activity and although needle electrodes are invasive, this method can be used to record the electrical activity of deep muscle. The use of needle electrodes is, however, uncommon outside of specialist research applications (Gleeson, 2001). Surface EMG is commonly used to quantify muscle co-contraction (van Dieen *et al*., 1996, Brown *et al*., 2007), identify muscle fatigue (Luttmann *et al*., 1996) and estimate strain according to previously defined reference levels (Attebrant *et al*., 1995, Doorenbosch *et al*., 2005).

The number and intensity of spikes in the EMG signal can be correlated to the force developed during muscle contraction and increased averaged EMG signals represent increased muscular contraction (Goebel, 2005). The amplitude of the EMG signal is dependent on a number of factors including electrode placement and size as well as the distance between the active muscle fibers. Consequently, the amplitude of the signal recorded will differ between individuals. To overcome this, EMG signals can be compared to maximum voluntary contractions (MVC) of the relevant muscles with EMG values expressed as a percentage of maximum (Goebel, 2005).

EMG, is a variable signal which is dependent on many factors including subcutaneous fat thickness, skin impedance and temperature, electrode size and placement as well as crosstalk from adjacent muscles. Normalisation of the EMG signal accounts for some of the inter-subject and inter-muscular EMG variability allowing comparison between subjects, different muscles or varying electrode sites on different days (Lehman and McGill, 1999). Normalization is a procedure whereby absolute EMG values are expressed as a percentage of a reference EMG recording obtained during a maximal or submaximal contraction test. The most widely used reference value is the maximal myoelectrical activity measured during maximal voluntary isometric contractions (MVC's) (Vera-Garcia *et al*., 2009).

SUMMARY

Although much effort has been placed into the prevention of low back disorders, these types of disorders are still common within the work environment (Davis and Marras, 2000). LBDs have been associated with the stability of the lumbar spine (Granata and Wilson, 2001; Anders *et al*., 2005) and therefore, in order to avoid injury to this region, spinal stability must be maintained throughout task completion. The trunk musculature plays a vital role in the maintenance of spinal stability through agonist and antagonist muscle activation. Although such muscle activity increases spinal stability, it results in increased spinal loading and may also impair trunk postural control (Lee *et al*., 2008).

Postural control is essential for the successful completion of all activities and consequently should be investigated within the context of realistic tasks. The muscle activity associated with the maintenance of postural equilibrium during load manipulation activities may add to total spinal loading and may also impair trunk movement. The magnitude of muscle activity increase may be influenced by both the degree of body movement and the mass of the external load. Therefore, understanding the influence of body movement and increased external load mass would provide valuable information which may be used to reduce the incidence of low back injury and pain in the working environment.

CHAPTER III
METHODOLOGY

INTRODUCTION

Spinal stability is an important factor in the prevention of low back disorders (LBD) and loss of spinal stability may result in spinal injury (Reeves *et al.,* 2007). Although activation of the spinal muscles increases spinal stability, it also results in increased spinal loading (Granata and Marras, 1995b). However, these conclusions are based predominantly on static models, whereas most industrial tasks require movements and associated muscle contraction (Granata and England, 2006) which therefore questions the use of these static models. Many guidelines and biomechanical models have been developed to decrease the risks associated with manual materials handling (MMH) and the development of LBDs. However, these guidelines do not distinguish between the muscle activity attributed to the external load and the underlying muscle activity which is used to maintain body equilibrium. In addition to this, it has been proposed that the presence of co-contraction during work-related tasks may cause guidelines to incorrectly predict the forces placed on the spine (Granata and Marras, 1995a; Gardner-Morse and Stokes, 1998). To the author's knowledge, the effect of load and trunk movement on muscular activity and body equilibrium has not yet been established and in addition to this, the individual effect of both load and degree of movement has not previously been quantified, hence the importance of this investigation.

Horak and MacPherson (1996) demonstrated that postural control is task dependent and even so, only a small number of studies have considered the influence of task demands on postural control (Prioli *et al.,* 2006). Additionally, few studies have investigated the postural control strategies during lifting or performing a manual handling task (Toussaint *et al.,* 1997a; Toussaint *et al.,* 1998; Oddsson *et al.,* 1999).

EXPERIMENTAL DESIGN

GENERAL EXPERIMENTAL CONCEPT

Although the control of body posture and the maintenance of upright stance is essential for both task completion and human movement (Horak *et al.,* 1997; Pollock *et al.,* 2000; Prioli *et al.,* 2006) it is typically not quantified in overall load analyses. Established ergonomic guidelines do not account for the increased strain placed on the body in instances which have a greater need for postural control (Kollmitzer, 2002). In such conditions, guidelines may need to be adapted to account for the postural control mechanisms which are necessary to maintain postural stability. As a result, this study aims to determine whether increased load and trunk movement affects the level of muscle activation necessary to maintain postural equilibrium during common industrially related tasks.

The study of balance control has largely focused on the recovery from single perturbations caused by an external force or volitional movement in which perturbations were inflicted and recovery strategies observed (e.g. Rietdyk *et al.,* 1999; Wilson *et al.,* 2006). Additionally, few studies have investigated balance control during continuous perturbations (Buchanan and Horak, 1999; Corna *et al.,* 1999; Akram *et al.,* 2008). Investigations regarding balance and postural control have tended to focus on the lower extremities and thus only limited information is available on the behaviour of the trunk during balance and postural control (Preuss and Fung, 2007). This is partly due to the fact that previous studies such as those performed by Horak and Nashner, 1986; Hughes *et al.,* 1995; Runge *et al.,* 1999 as well as Bothner and Jensen, 2001 have regarded the trunk as a single rigid structure whereas McGill *et al.* (1994) showed that small changes in trunk posture were achieved with the application of small moment of force which suggests that the trunk should not be considered as an inflexible column. A greater focus on the trunk and the reactions of the trunk muscles are, however, becoming more common.

This study proposes that when investigating postural control, posture and task characteristics should not be separated as they interact during task execution (Pollock *et al.,* 2000). Effective task completion is dependent on aspects such as the posture adopted, the movement of the centre of pressure (COP) beneath the feet and force production. These aspects in turn influence the way in which the task is performed and, ultimately, the strain placed on the individual (Figure 4). The mechanisms of postural control occur through closed loop feedback and in order for equilibrium to be maintained, postural control mechanisms are constantly updated and adapted (van der Krooij *et al.,* 2005). Hence, when investigating postural control during task completion it is necessary to investigate the strain placed on an individual without preventing feedback from occurring. Therefore this investigation analysed body sway and muscle recruitment patterns during repetitive load manipulation activities.

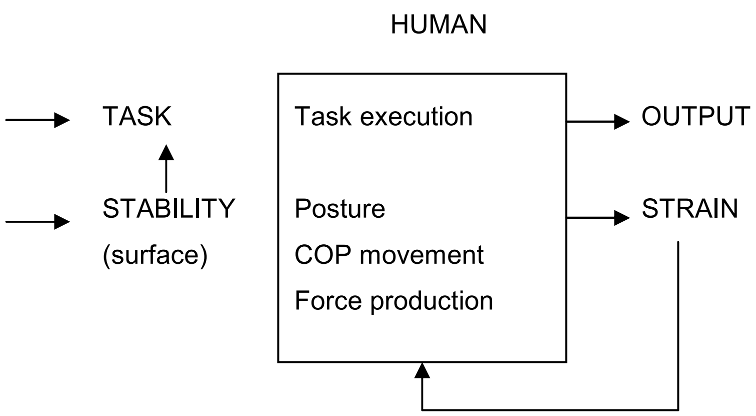


Figure 4: Feedback loop during task execution.

Task difficulty

Research has shown that increases in task difficulty result in a greater need for postural control (Prioli *et al.,* 2006). Task difficulty as well as the need for postural control could be influenced by altering the surface on which tasks are performed. A variety of surface conditions have been investigated including uneven (Menz *et al.,* 2003; Marigold and Patla, 2008), slanted, unstable (Radebold *et al.,* 2001; Silfies *et al.,* 2003) and rotating surfaces (Blackburn *et al.,* 2003; Akram *et al.,* 2008). Although information gained from altering support surface may be interesting, their applicability to an industrial setting is limited and subsequently was not included in the present investigation. Alterations to stance conditions have also been shown to increase task difficulty (Fujiwara *et al.,* 2008). Specifically it has been shown that stance width in a mediolateral direction has a great impact on postural stability (Day *et al.,* 1993; Henry *et al.,* 2001; Chiari *et al.,* 2002; Fujiwara *et al.,* 2008). Therefore, to allow for comparisons between conditions and to standardise task difficulty, stance width was kept constant (shoulder width apart) during this investigation.

Performing dual tasks such as answering arithmetic sums during task completion has also been shown to increase task difficulty and subsequently increases trunk sway (Verhoeff *et al.,* 2009). Although workers are often required to perform mental and physical tasks simultaneously, the effect of mental tasks was not the focus of this investigation and as such was not included in analysis. Altered visual input, for example performing tasks with closed eyes, has been shown to challenge the maintenance of equilibrium but the influence of vision on postural control has been well documented (Kuo *et al.,* 1998; Day *et al.,* 1993; Prioli *et al.,* 2006). In addition, it is uncommon for workers to have altered visual input during task completion and therefore visual input was not altered.

Increased velocity of movement is another factor which would alter stability demands of a task, requiring a greater degree of postural control to prevent loss of balance. Tasks involving greater trunk velocity have been shown to significantly increase the risk of low back injuries in the workplace (Norman *et al.,* 1998) and trunk stability has been estimated to decrease as the rate of flexion-extension increases (Granata and England, 2006). In addition, increased movement pace has been associated with an increase in muscle co-activation which ultimately leads to increased spinal loading (Davis and Marras, 2000). Although performing tasks at a variety of velocities would provide useful information, velocity of movement is not easily altered in industrial settings. Further, it would be difficult to provide industry with realistic and meaningful recommendations regarding movement speed as work pace is often determined by other factors such as conveyer belt speed. Therefore movement speed was kept constant throughout this investigation and subjects performed 1 lift and lower cycle every 6 seconds.

Postural equilibrium can also be challenged by the addition of an external load or by increasing the magnitude of a load (Commissaris and Toussaint, 1997; Li and Aruin, 2008). This study altered task difficulty by increasing the magnitude of the external load in both static and dynamic tasks. In such conditions it would be expected that the increased need to maintain postural equilibrium due to increased external load would result in greater body sway, indicated by the movement of the centre of pressure beneath the feet. This would be accompanied by an increase in muscle activity in order to maintain the centre of pressure (COP) within the base of support (BOS) to prevent falling. In this study, three varying loads (0.8kg, 1.6kg and 4kg) were used to assess the influence of increased load on body sway and trunk muscle activity.

Influence of load

Traditionally, research has focused on activities involving the lifting of heavy loads; however, the nature of work has changed and nowadays, workers involved in manufacturing are more often exposed to highly repetitive work involving less forceful motions (Punnett *et al.,* 1991). Accompanying this, there has been an increase in assembly jobs in which workers perform a variety of tasks in a work shift as opposed to the notion of one worker, one task (Marras *et al.,* 2009). As the work environment has changed, research should follow these trends and examine the associated changes in musculoskeletal loading (Westgaard and Winkel, 1997). Taking this into account, this research investigated four different load manipulation tasks performed under three relatively light loading conditions.

Trunk motion

Trunk movement can significantly alter the mechanical forces acting on the spine and consequently result in altered lumbar muscle activation (Dolan *et al.,* 1994; Dolan and Adams, 1998). Further, trunk motion plays an important role in postural control (Blackburn *et al.,* 2003). In order to accurately assess changes in postural equilibrium whole body movements employed during task execution should remain constant. Therefore tasks themselves should be the same but aspects of the task should be altered in order to challenge postural equilibrium. In order to assess the characteristics of the trunk under a variety of conditions, all tasks investigated involved trunk motion and body movement took place in all three anatomical planes of the body.

RESEARCH HYPOTHESIS

This research proposed that the muscle activity and movement of the centre of pressure attributed to the maintenance of postural equilibrium may place additional strain on the body in addition to that imposed by task demands. As the mechanisms of postural control may be affected by load magnitude as well as by the type of movement performed, the Null hypothesis states that the difference between the pattern of muscle recruitment and ground reaction forces of different loads will be proportional to the load difference (load component of postural control). Additionally it is hypothesised that the difference in muscle activity and ground reaction forces between static and dynamic tasks performed under the same loading condition is the same when loads are increased (movement component of postural control). Further, it is hypothesised that increased external load and increased trunk movement will not affect psychophysical responses during task completion.

DESIGN MATRIX

This research proposes that the effort necessary to maintain postural equilibrium is influenced by two variables, namely the magnitude of the external load and the degree of movement which occurs during task completion. In order to isolate these components during task execution, it is necessary to investigate the muscle activity and centre of pressure movement associated with both static and dynamic tasks, as well as the changes in responses while performing tasks with different external loads. Direct measurement of the mechanisms of postural control is complex as they underly all movements (Houdijk *et al.,* 2009). Therefore, the movement of the centre of pressure and the muscle activity associated with different tasks were used as indicators of the muscle activity necessary to maintain postural equilibrium. To the author's knowledge there is limited research which aims to separate the aspects of movement and load and their effect on both body sway and muscle activity during the completion of static and dynamic tasks.

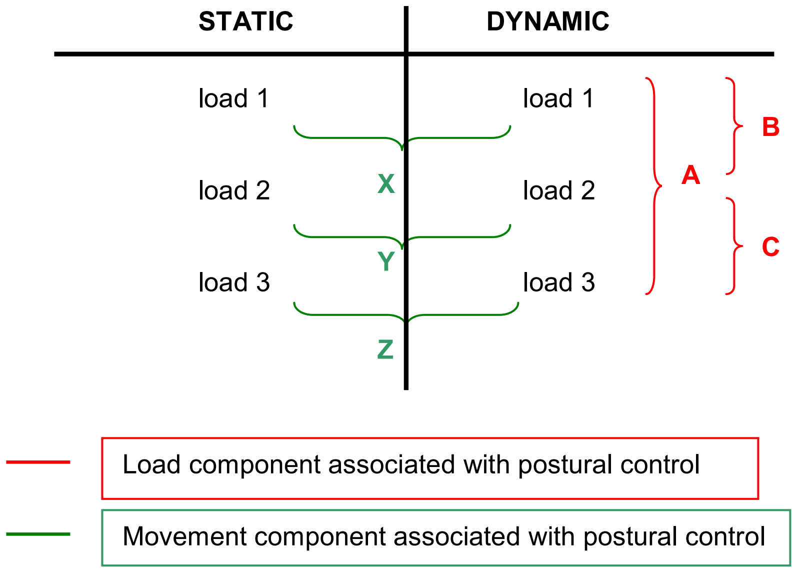


Figure 5: Basic experimental concept and design indicating the proposed load component and movement component of postural control.

Differential analysis was necessary in order to isolate the effect of load and movement. Total response elicited during task completion includes biomechanical factors such as load mass and lever length; the effort required to maintain posture; as well as the effort required to maintain postural equilibrium. Differential analysis allowed the biomechanical factors, the effort to maintain posture and other factors to remain constant as differences between responses of one subject are calculated. Further, using this type of analysis the proposed “movement component” and “load component” of postural control can be calculated.

With reference to Figure 5 the proposed “load component” and “movement component” of postural control can be separated by calculating the difference in responses elicited by each condition. Firstly, the difference in responses between the three loading conditions will provide an indication of the “load component” of postural control. Secondly, the difference between the static and dynamic conditions will show the “movement component” of postural control. The differences in responses were compared, thereby avoiding direct comparisons between static and dynamic tasks.

Load component of postural control

The total response during task execution will include the effort required to maintain postural equilibrium as well as the influence of task factors. In addition, the maintenance of equilibrium is influenced by biomechanical factors such as body mass, lever length and other anthropometric data. The influence of biomechanical factors is, however, constant over both static and dynamic conditions and between subjects. Therefore, it can be proposed that the total response elicited by an individual will be influenced by body movement, external load, postural control and biomechanical factors.

With reference to Figure 5, responses of each loading condition can be compared under both static and dynamic conditions.

Therefore:

Where : R = response

$R\ load\ 1 = R_{movement\ component} + R_{load\ component\ 1} + R_{biomechanical\ factors} + R_{posture} + R_{load\ 1}$

$R\ load\ 2 = R_{movement\ component} + R_{load\ component\ 2} + R_{biomechanical\ factors} + R_{posture} + R_{load\ 2}$

$R\ load\ 3 = R_{movement\ component} + R_{load\ component\ 3} + R_{biomechanical\ factors} + R_{posture} + R_{load\ 3}$

As the biomechanical factors remain constant (due to the responses of the same subject) their effect cancels out and therefore need not be considered in analysis. In order to isolate the effect of load, the difference in response between the loading conditions must be compared.

Therefore:

Where : R = response

$R\ load\ 3 – R\ load1 = R_{load\ component\ load\ 3} – R_{load\ component\ load\ 1} + 4\ R_{load\ 1}$

$R\ load\ 2 – R\ load\ 1 = R_{load\ component\ load\ 2} – R_{load\ component\ load\ 1} + R_{load\ 1}$

$R\ load\ 3 – R\ load\ 2 = R_{load\ component\ load\ 3} – R_{load\ component\ load\ 2} + 3\ R_{load\ 1}$

The load magnitude increased by different increments between the loading conditions. The load doubles in magnitude between the first and second load (0.8kg and 1.6kg) and increases by five times between load 1 and load 3 (0.8kg and 4kg). Therefore if the load component of postural control is linear, the responses associated with load 2 (1.6kg) will be double that of load 1 (0.8kg). Further, responses associated with load 3 (4kg) will be five times that of load 1 (0.8kg).

Where : R = response

$R_{\text{load component 1.6kg}} = 2 \times R_{\text{load component 0.8kg}}$

$R_{\text{load component 4.0kg}} = 5 \times R_{\text{load component 0.8kg}}$

If the above holds true:

Where : R = response

$R4.0kg – R0.8 = 4\ R_{\text{load component 0.8kg}} + 4\ R_{\text{load 0.8kg}}$

$R1.6kg – R0.8kg = R_{\text{load component 0.8kg}} + R_{\text{load 0.8kg}}$

$R4.0kg – R1.6kg = 3\ R_{\text{load component 0.8kg}} + 3\ R_{\text{load 0.8kg}}$

A: $R4.0kg – R0.8kg = 4\ (R_{\text{load component 0.8kh}} + R_{\text{load 0.8kg}})$

B: $R1.6kg – R0.8kg = 1\ (R_{\text{load component 0.8kg}} + R_{\text{load 0.8kg}})$

C: $R4.0kg – R1.6kg = 3\ (R_{\text{load component 0.8kg}} + R_{\text{load 0.8kg}})$

Therefore, referring to Figure 5, if:

B = ¼ A	the question holds true and load component of postural control is linear
B ≠ ¼ A	the load component of postural control is not linear
C = ⅓ A	the question holds true and load component of postural control is linear
C ≠ ⅓ A	the load component of postural control is not linear

Statistical hypothesis - load component of postural control

Hypothesis 1: The null hypothesis states that the difference in responses between load 1 and 3 (indicated by A on the figure) will be three times the magnitude of the difference in responses between load 2 and 3 (C) and be four times the magnitude of the difference in responses between loads 1 and 2 (B). This is expected for all tasks performed.

H_o: A = 4B and A = 3C
H_a: A ≠ 4B and A ≠ 3C

Where : A, B and C are loading conditions

: responses = electrical activity of selected trunk muscles, ground reaction forces and psychophysical responses

Movement component of postural control

In order to isolate the muscle activity and ground reaction forces associated with body movement, the difference in responses between the static and dynamic conditions under the same loading conditions can be compared (indicated by # in Figure 5:46). The difference in responses between the static and dynamic conditions would provide information regarding the "movement component" of postural control. An increase in muscle activity and body sway would indicate a greater need for postural control, whereas a decrease could be indicative of movement efficiency which may occur in dynamic tasks. It is important to note that the differences in responses between the static and dynamic tasks are compared and that static and dynamic tasks are not directly compared to each other.

Statistical Hypotheses – movement component of postural control

Hypothesis 2: The null hypothesis states that the difference in responses between the static and dynamic condition for all tasks, performed with the same external load mass will be the same.

H_o: X = Y = Z

H_a: X ≠ Y ≠ Z

Where : X, Y and Z are the difference in responses between static and dynamic conditions

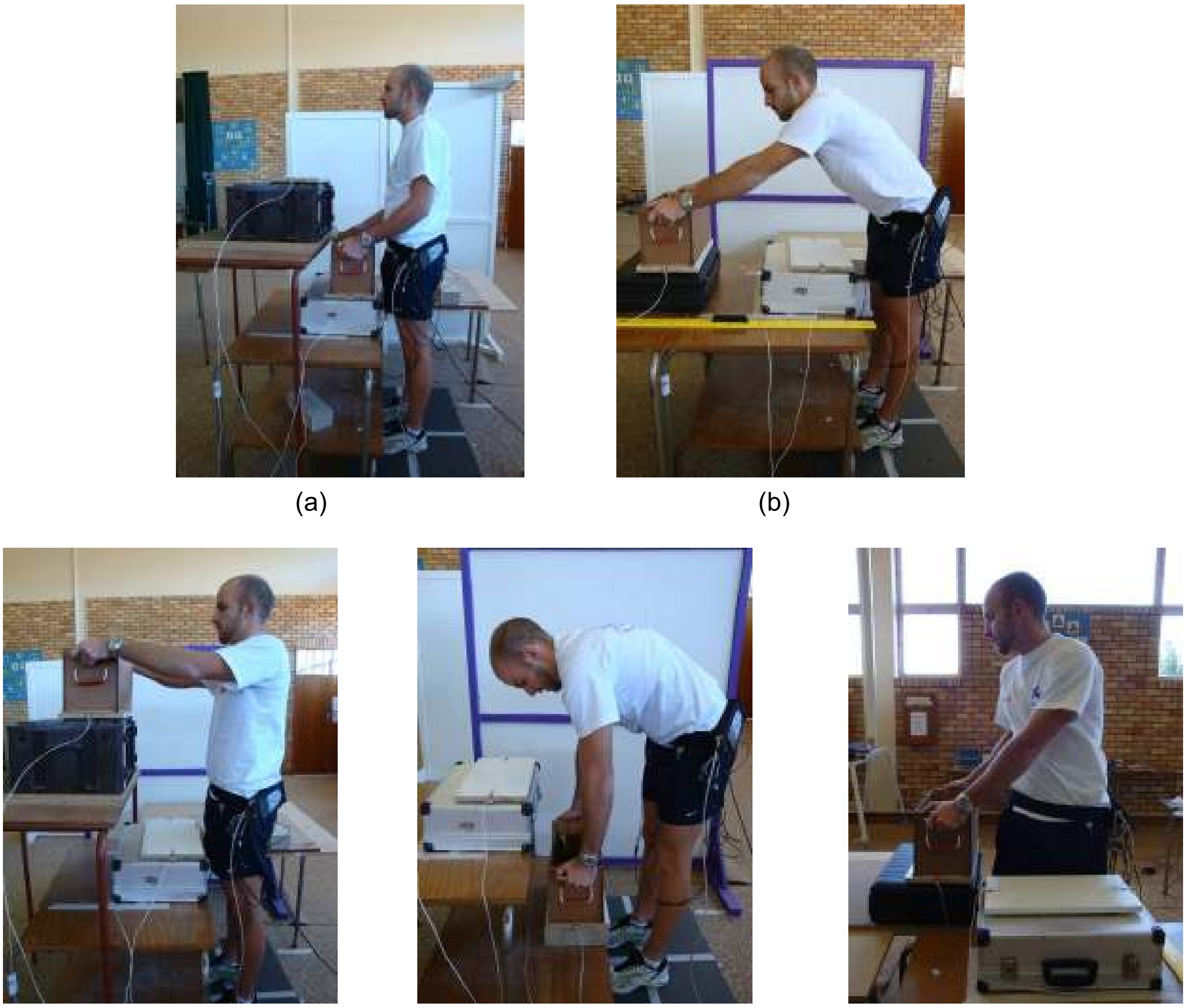
: responses = electrical activity of selected trunk muscles, ground reaction forces and psychophysical responses

Tasks investigated

Four load manipulation tasks under both symmetrical and asymmetrical conditions were chosen for this investigation. These types of tasks allow for stationary foot placement and would therefore comply with the size of the force platform. The tasks selected incorporated movement along the three cardinal planes and all of the dynamic tasks involved trunk movement. The force produced during task completion was in the same direction as trunk movement as it is unlikely that individuals would adopt a movement pattern such that trunk movement would be in the opposite direction to force production. A description of each of the four tasks can be seen in Table I. The corresponding static position for each of the selected tasks was set at the end position of the particular task and not a neutral standing position as the end position would elicit the greatest amount of strain on subjects. The start position of all of the tasks remained constant and can be seen in Figure 6 together with the end position associated with each task. The selection of these tasks allowed for a broad analysis into the maintenance of postural equilibrium during a variety of work-related tasks.

Table I: Description of static and dynamic conditions of the tasks investigated.

TASK NAME	DYNAMIC	STATIC
1."Hip knee"	Moving the load between hip height and knee height	Box positioned at knee height
2."Hip shoulder"	Moving the load between hip height and shoulder height	Box positioned at shoulder height
3. "Hip twist"	Twisting from midline to 90° on dominant side at hip height	Box positioned 90° from midline on dominant side
4. "Hip reach"	Transferring the load at hip height from 15cm away to an arm length from the body	Box positioned at hip height an arm length away



(a) (b)

(c) (d) (e)

Figure 6: Experimental setup showing (a) start position and end positions for "hip- reach" (b), "hip- shoulder" (c), "hip-knee" (d) and "hip-twist (e) tasks.

For each task, subjects were required to perform a dynamic and static condition. To complete the dynamic condition associated with each task, subjects were required to transfer the box (390mm x 190mm x 210mm) as described in Table I, within the demarcated area. Movement pace was set at one transfer every three seconds and movement occurred for the duration of one minute. Therefore, the box was moved from, and back to its original position in six seconds and ten cycles performed in each dynamic condition. Each task was completed under three loading conditions such that the difference between the conditions, and the effect of load, could be calculated. A modified box was used for the lightest loading condition and had a mass of 0.8kg. As the maximum load was set at four kilograms, the third loading condition was set midway between the two loads at 1.6kg. The boxes used in this investigation can be seen in Figure 7.

Figure 7: Modified box constituting the lightest loading condition (right) and box used for additional two loading conditions (left).

For each task, the box was placed 150mm away from the subjects' ankle bones as this is the optimal horizontal component proposed by Waters *et al.* (1993). All of the task parameters were altered according to each subject's anthropometric data such that the handles of the box were positioned at the relevant heights required by the respective tasks. For the "hip-reach" task, the box was placed an arm's length from the subject at hip height.

Dependent variables

The muscle activity used to perform each of the tasks was considered to be a necessary measurement, as the risk of work-related musculoskeletal disorder development is increased when high levels of muscle activity are sustained (Straker and Mekhora, 2000). In addition, the muscular response to postural perturbations can be quantified using electromyography (Visser *et al.,* 2008). In order to select the relevant muscles for analysis, previous research was studied and the most common muscles used were selected for pilot investigation. The muscles selected for investigation included the right and left lumbar erector spinae (location of largest muscle mass approximately 40mm from midline of the spine), right and left thoracic erector spinae (location of largest muscle mass at the level of the last rib) right and left internal oblique (40mm above the ilium in the lumbar triangle at an angle of 45° to the midline of the spine), external oblique (100mm from the midline of the abdomen and 40mm above the ilium at an angle of 45°), rectus abdominis (30mm from the midline of the abdomen 20mm above the umbilicus) and latissimus dorsi muscles (most lateral portion of the muscle at the level of T9) (Mirka and Marras, 1993). Illustration of electrode placement can be seen in Figure 8.

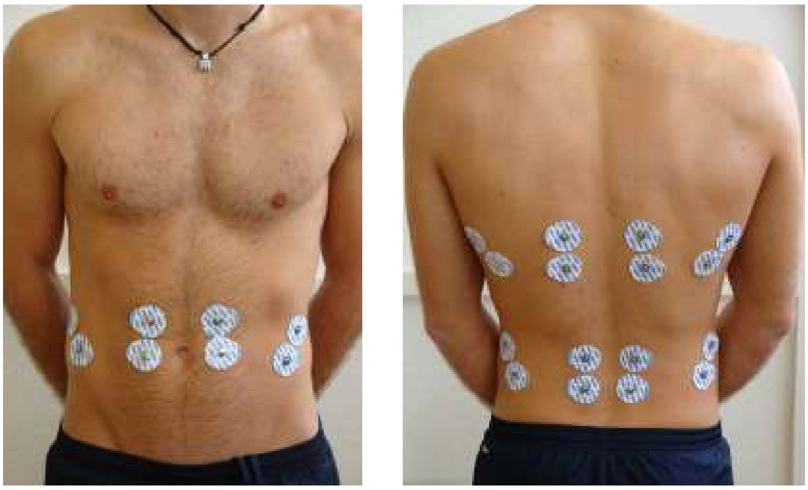


Figure 8: Electrode placement on anterior and posterior sides of the body.

Information regarding the ground reaction forces occurring during each task was also identified as a valuable dependent variable. Measurement of the movement of the centre of pressure (COP) also referred to as body sway, can be analysed through the use of a force platform. Body sway, which occurs in the horizontal plane, contains valid information regarding postural stability (Karlsson and Frykberg, 2000). Information regarding the movement of the centre of pressure during the investigated tasks would provide valuable insight into the muscle activity involved in maintaining postural equilibrium while performing common work-related activities.

Taking a holistic approach is important when evaluating human effort (Charteris *et al.,* 1976). Therefore, in addition to the biomechanical measure of the analysis of the movement of the COP and the physiological information gathered using electromyography (EMG), psychophysical assessments were also performed using the Body Discomfort Scale and Map developed by Corlett and Bishop (1976).

Pilot studies

Preliminary studies were conducted prior to the experimental investigation in the Human Kinetics and Ergonomics Department of Rhodes University. These studies aimed to establish acceptable external loads to be manipulated, the correct positioning of the electrodes, the selection of the most relevant muscles, movement speed and to establish which aspects of the investigation needing to be controlled.

The mass of the loads was determined by preliminary investigations. Two female subjects who did not partake in regular resistance exercise were used to determine the magnitude of the maximum loading condition. Subjects who did not participate in regular resistance exercise were chosen as the maximum load should be small enough such that the weakest subjects would be able to complete the heaviest loading condition. The subjects were required to complete the static and dynamic tasks under a variety of loading conditions. It was concluded that in order for subjects to maintain the static postures, a load of no greater than four kilograms should be used.

A variety of lifting frequencies ranging from eight to twelve lifts per minute were investigated to determine a comfortable lifting pace. During this investigation, subjects were asked to identify a lifting pace at which lifts could be performed comfortably without having a long rest period between lifts. Subjective feedback revealed that ten lifts per minute or one lift and lower every six seconds was optimal. In addition, these investigations showed that it was necessary to give instruction to subject to initiate lift and replace phases. As such, a metronome was used to control lifting pace and was set at one beat every three seconds so that a complete lift cycle was completed in six seconds.

Preliminary investigations were performed on three subjects to determine the most appropriate muscles to investigate as well as appropriate electrode placement. As this research was focused on the role of the trunk muscles in maintaining postural equilibrium, all superficial trunk muscles were considered for investigation. However, the availability of surface EMG excluded the analysis of muscles located deep within the muscular system. Pilot studies concluded that the multifidus muscle should be excluded from investigation due to its location and the difficulty in measurement with surface electromyography. Although the rectus abdominis muscle showed low levels of activation during pilot testing, the muscle was included in analyses as results from this muscle may infer activity of the transverse abdominis muscle which plays an important role in spine stabilisation.

As one of the tasks chosen involved twisting, six pairs of trunk muscles (both right and left sides) were investigated. These included the erector spinae (thoracic and lumbar portions), internal oblique, external oblique, rectus abdominis and latissimus dorsi muscles. These muscles were chosen due to their activation levels throughout the tasks as well as their superficial location in the muscular system.

MEASUREMENT AND EQUIPMENT

Anthropometric and Demographic data

Anthropometric data was collected from all subjects as task parameters were adjusted according to this information. This ensured that all subjects were placed under the same degree of strain during each of the tasks and also allowed for greater standardisation in the investigation, such that differences in results could be attributed to the task and not to subject variability. A Takei anthropometer was used to measure standing elbow height (floor to olecranon process), standing shoulder height (floor to acromion process), standing femur height (floor to head of femur) standing knee height (floor to centre of patella) and arm length (acromion process to the tip of the longest finger). Stature was measured to the nearest millimetre (mm) using a Harpenden stadiometer and mass was measured using a Toledo scale to the nearest 0.1kg. For standardisation purposes, anthropometric measurements were taken on the right-hand side of the body. Demographic data collected included age and sex. Anthropometric and demographic data can be seen in Table II:58.

Physiological measure - Electromyography

The activity of the upper and lower erector spinae, internal oblique, external oblique, rectus abdominis and latissimus dorsi muscles were measured bilaterally using surface electromyography (Muscle Tester Mega ME6000P16 Mega Electronics Ltd, Finland). The electrodes used were disposable, pre-filled Silver-Silver Chloride electrodes and were positioned on the belly of the muscle with a third electrode acting as a neutral electrode on inactive muscle. Data was normalised using maximum voluntary contractions of the relevant muscles. Each maximal contraction occurred against an external resistance provided by the researcher and was held for 5 seconds with 3 seconds used for analysis. Data was transmitted to a laptop computer and was processed using the MegaWin 2.4 software package.

Biomechanical measure – Ground reaction forces

A Bertec force platform was used in this investigation to measure the ground reaction forces and the movement of the centre of pressure during task completion. Three orthogonal forces and the moments about each axis were measured by the force plate by using strain-gaged load transducers (Bertec Corporation, 2008). This signal was then amplified and was connected to a computer using a USB connection and was processed using Bertec software. The force plate and the orthogonal planes can be seen in Figure 9.

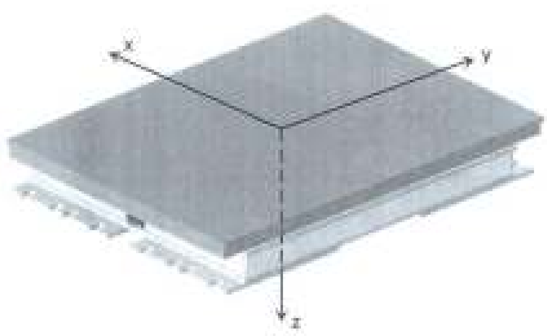


Figure 9: Bertec Force Plate indicating orthogonal planes (Bertec Corporation, 2008).

Psychophysical measure - Body Discomfort Scale and Map

The Body Discomfort Scale and Map by Corlett and Bishop (1976) divides the body into 22 regions, distinguishing between the anterior and posterior sides of the body. This allows discomfort to be accurately assigned to a specific region or body part. Areas experiencing discomfort are ranked using a Likert scale of one to ten with one representing no discomfort and ten representing extreme discomfort. It should be noted that although this method provides quantitative data, it is a subjective rating of discomfort and could vary between individuals. Subjects were requested to identify and rate body areas experiencing discomfort at the end of each condition. The Body Discomfort Scale and Map can be seen in Figure 10.

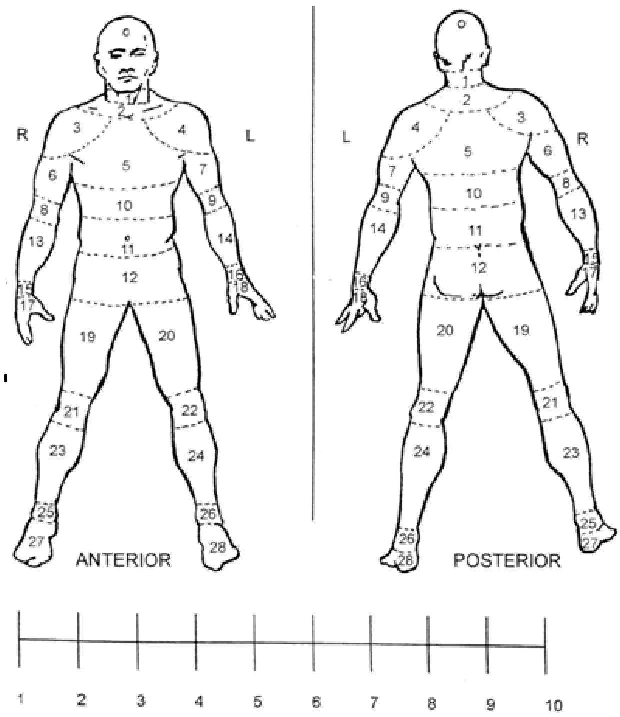


Figure 10: Body Discomfort Scale and Map (Corlett and Bishop, 1976).

EXPERIMENTAL PROCEDURE

Experimentation was conducted in the Ergonomics Laboratory of Rhodes University Human Kinetics and Ergonomics (HKE) Department. Subjects were required to participate in one testing session during which 24 conditions (four tasks, two movement conditions, three loading conditions) were assessed as shown in Table I (p.51) and further illustrated in Figure 6 (p.51). The order in which the 24 conditions were performed by each subject was determined prior to the testing session through permutation. This ensured that differences found in responses could be attributed to changes in the tasks investigated and not to familiarisation or fatigue.

The study involved 48 subjects (24 females and 24 males) from the Rhodes University population who volunteered to participate in the investigation. Stature was not restricted as the task setup was altered according to each of the subject's anthropometric data. All subjects partook in regular physical activity (three times per week) and had no history of low back injury or low back pain. In addition, none of the subjects had any professional experience in manual materials handling.

Table II: Anthropometric data for all subjects (n=48).
(mean with standard deviation in brackets)

	MALES	**FEMALES**
Age (Years)	22 (2.34) 10.74%	21 (1.76) 8.35%
Stature (mm)	1777 (56.93) 3.2%	1669 (63.84) 3.82%
Mass (kg)	74.4 (8.26) 11.11%	60 (8.62) 14.38%
Knee height (mm)	500 (46.54) 9.29%	466 (49.48) 10.62%
Femur height (mm)	964 (50.44) 5.23%	892 (62) 6.95%
Arm length (mm)	796 (59.48) 7.47%	730 (57.88) 7.93%
Acromion height (mm)	1435 (101.31) 7.06%	1356 (62.86) 4.64%
Standing elbow height (mm)	1069 (57.25) 5.36%	996 (48.54) 4.87%

% refers to coefficient of variation

Subjects were tested individually during the experimental session. On arrival, the aim of the research and the tasks investigated were explained. The measurement devices (EMG and force platform) were briefly explained again to the subjects such that the subjects had a better understanding of the measurements taking place. Questions were addressed and informed consent signed. Subjects were then allowed to practice each of the tasks in order to allow subjects to familiarise themselves with the conditions. Following this, anthropometric data were recorded. The locations for electrode placement were cleaned with Milton fluid and shaved, if necessary, to ensure good connectivity. Electrodes were placed on the skin in the relevant positions of the 6 pairs of muscles tested (Figure 8:53). Following this the EMG device was connected.

Prior to the start of testing, subjects were required to perform maximum voluntary contractions (MVCs) for each of the selected muscles. These data were used to determine the percentage of maximum at which subjects were working during task completion. To perform MVCs, subjects were required to adopt specific postures (Figure 11) and were asked to exert maximal force against resistance provided by the researcher. MVC data were collected for the duration of five seconds, and each MVC was performed twice. A minimum of 30 seconds rest was provided between each of the contractions to prevent fatigue. According to literature, optimal duration for MVCs is between two and six seconds (Visser *et al.,* 2004, Escorpizo and Moore, 2006) hence five seconds were determined sufficient for this research.

Figure 11: Postures adopted for MVC calibration of the lumbar and thoracic erector spinae, internal and external oblique, latissimus dorsi and rectus abdominis muscles (Kendall *et al.,* 1993).

Following MVC completion, the workstation was adjusted according to the subjects' anthropometric data and the task which was to be performed. Subjects were then instructed to stand in the middle of the force platform, feet positioned shoulder width apart and foot position was marked, which ensured that subjects stood in the same position during experimentation. Once subjects were standing in the correct position on the force plate and the workstation was adjusted, testing began.

EMG and force plate data were collected throughout the testing session. Data from switches attached to the EMG device indicated the beginning and end of periods of activity. This information was used to aid accurate data analysis as the start and end of each movement phase could be seen with the activation or deactivation of the switches. A one minute rest period was provided in-between tasks in order to avoid fatigue. Body Discomfort was recorded during the rest period provided after the completion of each condition.

STATISTICAL ANALYSIS

Muscle activation and ground reaction force pattern was analysed during each task. Movement pattern was analysed in relative timing such that each lift and lower cycle represented 100% of time taken to complete the task. Lift and lower movements were analysed separately for every subject and for all tasks investigated. Each movement phase was separated into 12 intervals of equal size. Data in each interval was averaged to create a 12 data point pattern for each movement phase for each subject for all tasks.

For statistical analysis, manipulated data (see Figure 5:46) was analysed separately for each task (“hip knee”, “hip shoulder”, “hip twist” and “hip reach”), and for each muscle/ ground reaction force, using a two factorial analysis of variance (ANOVA) (response, condition). Significance level was set at 0.05.

CHAPTER IV
RESULTS

INTRODUCTION

The effect of load mass and body movement on muscle activity, ground reaction forces and psychophysical responses was studied in a laboratory setting. Four load manipulation tasks as described in Table I were investigated under static and dynamic conditions and under three loading conditions (0.8kg, 1.6kg and 4kg). Electromyography (EMG) of six pairs of trunk muscles, namely the thoracic and lumbar erector spinae, internal oblique, external oblique, rectus abdominis and latissimus dorsi muscles were investigated throughout task completion. Additionally, ground reaction forces were measured through the use of a force platform during the testing session. After the completion of each condition, subjects were required to rate postural stress using the Body Discomfort Map and Scale developed by Corlett and Bishop (1976).

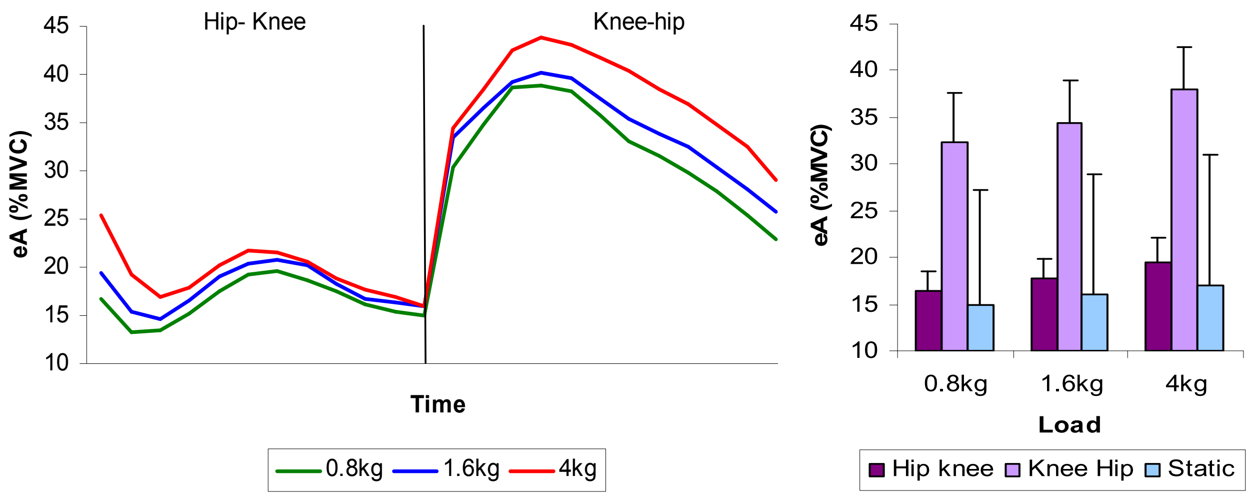
In traditional load analyses, movements are typically analysed by averaging data over a number of repetitions. This method of analysis provides information regarding the underlying average movement pattern but does not account for the muscular effort related to movement variability and co-contraction (Houdijk *et al.,* 2009). Further, averaged data does not reveal the extra muscular effort associated with balance control (Houdijk *et al.,* 2009). Therefore, pattern analysis was used in this investigation and data was not averaged over a number of repetitions. Pattern analysis of muscle activity and ground reaction forces was performed for each subject such that one representative lift and lower cycle was determined for each task. Patterns were generated by analysing twenty four data points in one lift and lower cycle in a period of 6 seconds. Differential analysis was used to determine the proposed “load component” and “movement component” of postural control as illustrated in Figure 5 (p.46).

Prior to data manipulation, pattern of muscle activity and ground reaction forces of each task was determined Pattern of muscle activity and ground reaction forces associated with the “hip knee” task are illustrated below. The same analysis was performed for all tasks investigated and can be seen in Appendix C (p.163).

Muscle activity pattern associated with the "hip knee" task

As the "hip knee" task was symmetrical in nature, pattern of muscle activation of the right and left muscles was similar. Consequently, only right hand side muscles are illustrated. Pattern of muscle activation of the left muscles during the "hip knee" task are displayed in Appendix C.

For all muscles investigated, the lowering motion (hip-knee) of the task resulted in smaller muscle activity levels compared to the lifting motion (knee–hip) under all investigated loading conditions. This can most likely be attributed to moving the load in the same direction as gravity. Increased load magnitude resulted in increased average muscle activity for the hip-knee, knee-hip and static conditions in all muscles except the right internal oblique and right latissimus dorsi muscles in which average muscle activity decreased in the 1.6kg loading condition. The knee-hip movement phase elicited the highest average muscle activity under all loading conditions in all investigated muscles. Additionally, average muscle activity of the static condition (box held at knee height) was lower than the average muscle activity of the hip-knee and knee-hip movement phases.



(A) (B)

Figure 12: Muscle activation pattern (A) and average activation (B) of the right lumbar erector spinae muscle during the "hip knee" task. [1]

[1] Note that the scale of muscle activation (%MVC) is not constant for each muscle due to differences in observed muscle activity levels.

The pattern of muscle activity of the right lumbar erector spinae remained a similar shape but increased in amplitude with increased external load (Figure 12A). The pattern of muscle activity was greatest during the 4kg loading condition and peaked at 25.3% MVC during the hip-knee movement directly after load pick up and 43.7% MVC midway during the knee-hip movement. Greatest average muscle activity occurred in the knee-hip movement phase under the 4kg condition and had an average muscle activity of 38% MVC (Figure 12B).

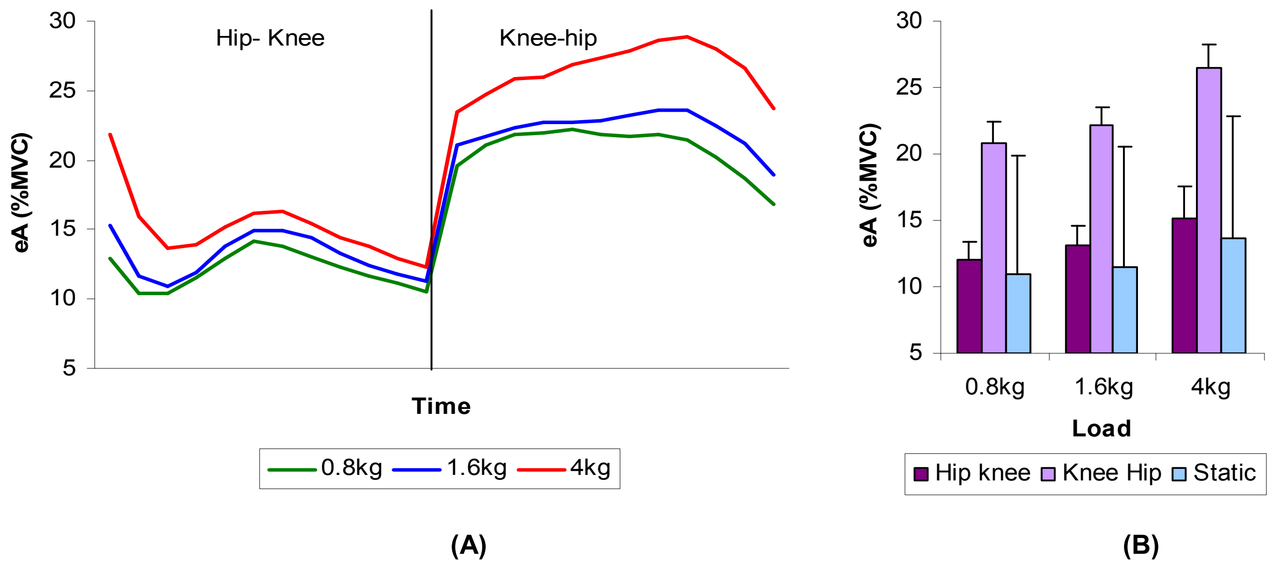
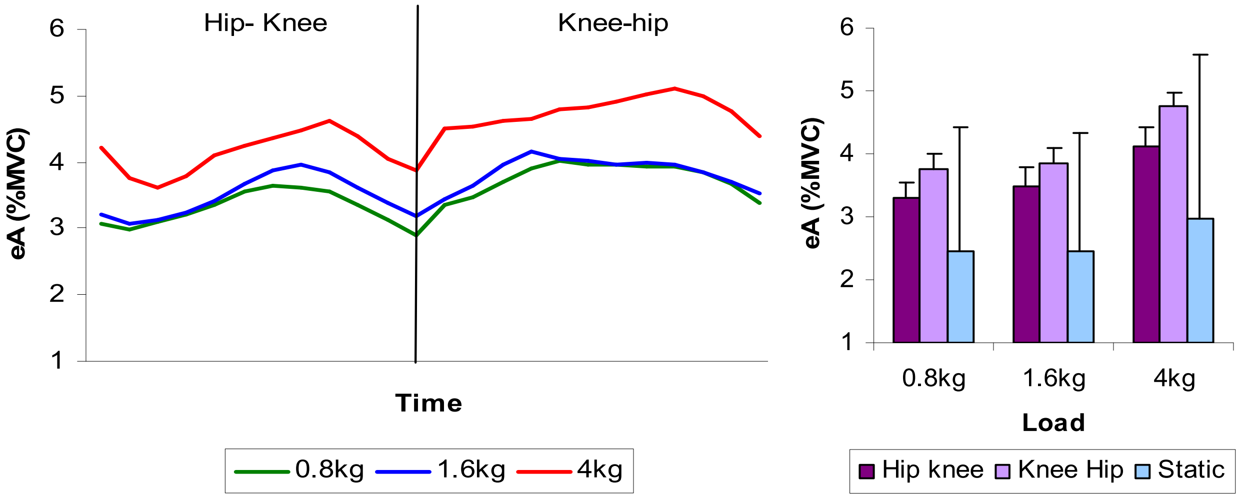


Figure 13: Muscle activation pattern (A) and average activation (B) of the right thoracic erector spinae muscle during the "hip knee" task.

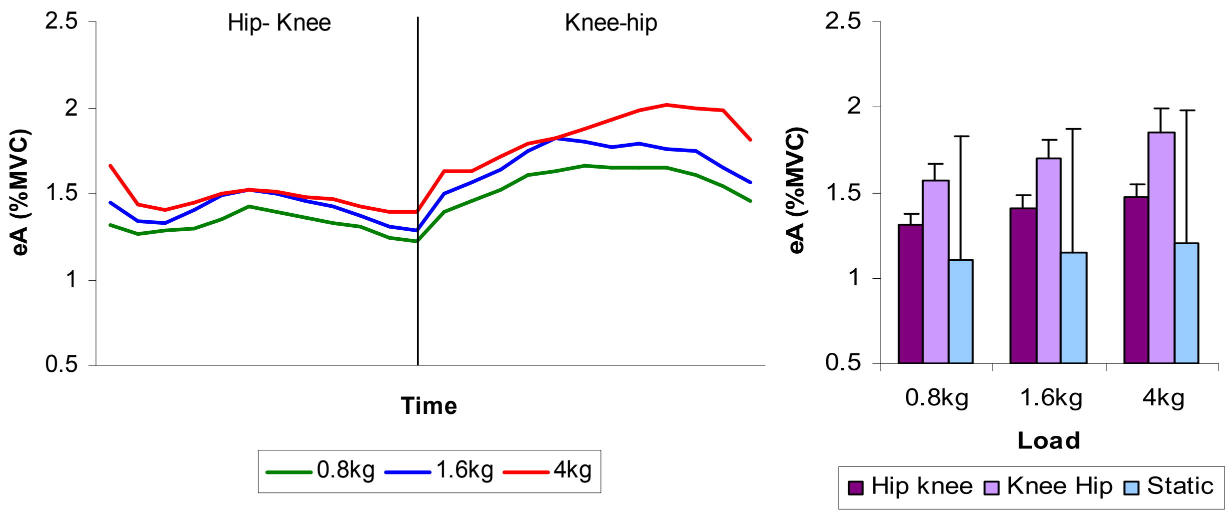
The pattern of muscle activity of the thoracic erector spinae muscle followed a similar trend to the responses of the lower erector spinae. Muscle activation was highest during the 4kg loading condition and peaked at 21.8% MVC during the hip-knee movement and 28.8% MVC during the knee-hip movement (Figure 13A). Average muscle activity peaked at 26.4% MVC during the knee-hip movement phase of the 4kg loading condition (Figure 13B).



(A) (B)

Figure 14: Muscle activation pattern (A) and average activation (B) of the right external oblique muscle during the "hip knee" task.

The 0.8kg and 1.6kg loading conditions elicited similar muscle activation pattern and average activation level of the right external oblique during both the hip-knee and knee-hip movements (Figure 14A). The 4kg loading condition produced a similar pattern of response but muscle activation levels were higher, peaking at 4.6% MVC during the hip-knee movement and 4.9% MVC during the knee-hip movement. The hip-knee and knee-hip movement phases resulted in similar average muscle activation levels and differed by 0.63% MVC, 0.38% MVC and 0.47% MVC for the 0.8kg, 1.6kg and 4kg condition respectively (Figure 14B).



(A) (B)

Figure 15: Muscle activation pattern (A) and average activation (B) of the right rectus abdominis muscle during the "hip knee" task.

The level of muscle activation of the rectus abdominis muscle increased under greater external loads. Difference in average muscle activation between the 4kg and 0.8kg loading condition was 0.15% MVC for the hip-knee movement phase and 0.28% MVC for the knee-hip movement phase (Figure 15B). Average static muscle activation (Figure 15B) was lower than the pattern of movement (Figure 15A) during all loading conditions and during both movement phases.

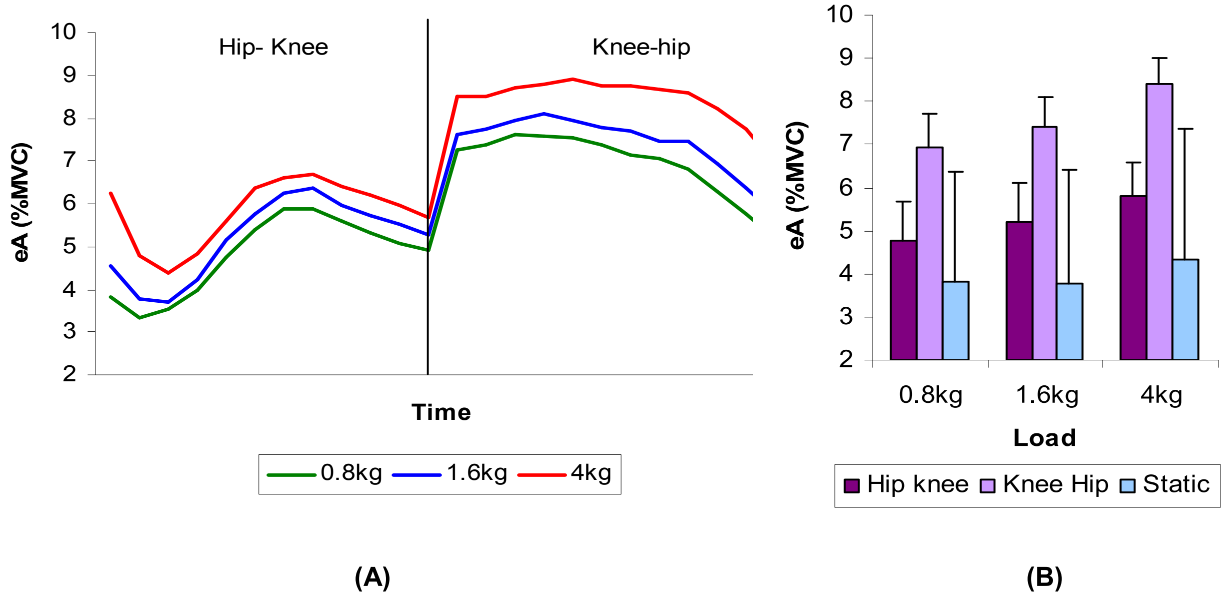
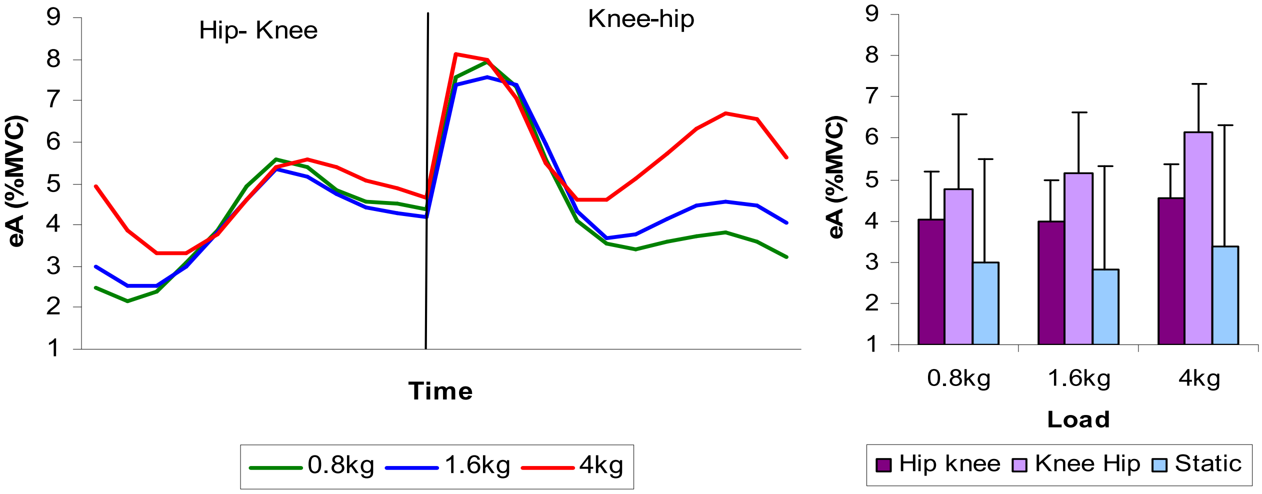


Figure 16: Muscle activation pattern (A) and average activation (B) of the right internal oblique muscle during the "hip knee" task.

The pattern of muscle activity of the right internal oblique remained a similar shape and increased under greater loading conditions (Figure 16A). Muscle activity increased sharply at the start of the lifting (knee-hip) phase and increased by 2.37% MVC, 2.34% MVC and 2.84% MVC for the 0.8kg, 1.6kg and 4kg loads respectively (Figure 16A). Average static muscle activity decreased in the 1.6kg loading condition by 0.02% MVC and increased in the 4kg loading condition to 4.3% MVC (Figure 16B).



(A) **(B)**

Figure 17: Muscle activation pattern (A) and average activation (B) of the right latissimus dorsi muscle during the “hip knee” task.

The muscle activation pattern of the right latissimus dorsi followed a similar trend under all loading conditions with load having the greatest effect on muscle activation pattern at the beginning of the hip-knee movement phase and at the end of the knee-hip movement phase (Figure 17A). The difference in muscle activity between the 0.8kg and 4kg condition was 2.4% MVC at the onset of the hip-knee movement phase and was 2.9% MVC nearing the end of the knee-hip phase. During the middle phase of movement, level of muscle activation was similar under all loading conditions. Muscle activity associated with the 0.8kg condition exceeded that of the 1.6kg condition during the hip-knee and knee-hip movement phases (Figure 17A). Average muscle activity during the static condition decreased from the 0.8kg condition to the 1.6kg condition and was greatest during the 4kg loading condition (Figure 17B).

Ground reaction force pattern of the “hip knee” task

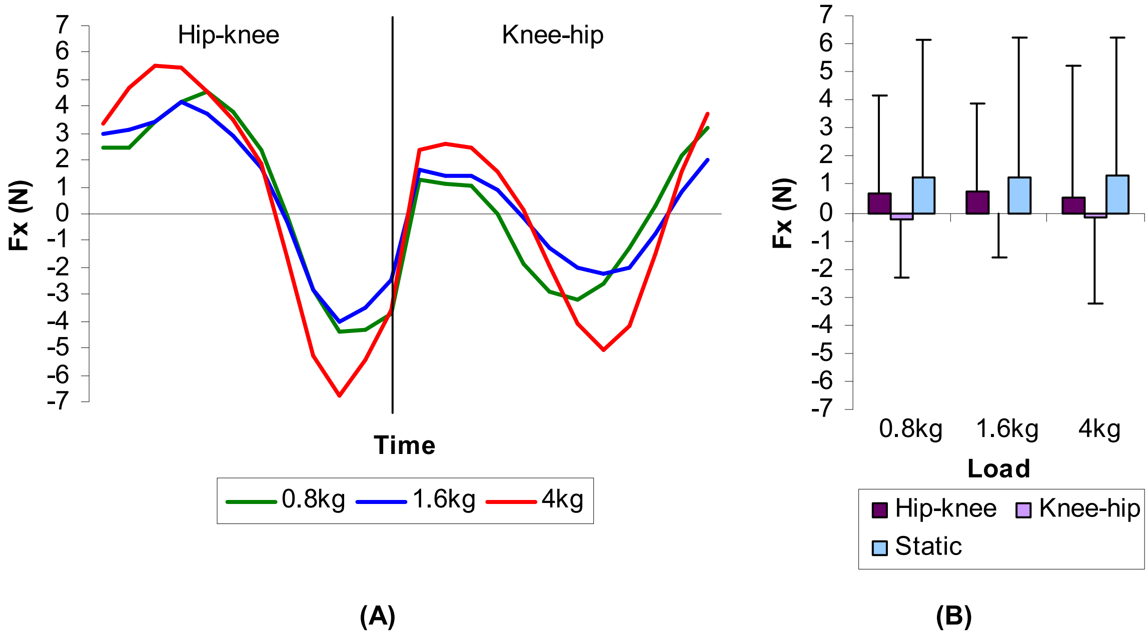
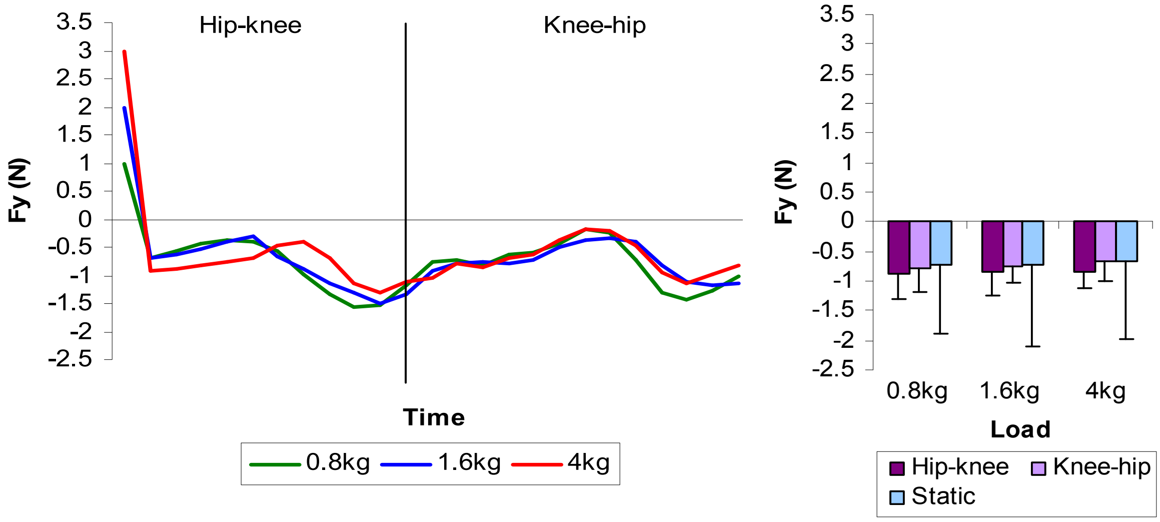


Figure 18: Pattern (A) and average (B) sagittal ground reaction forces under each loading condition during the “hip knee” task.

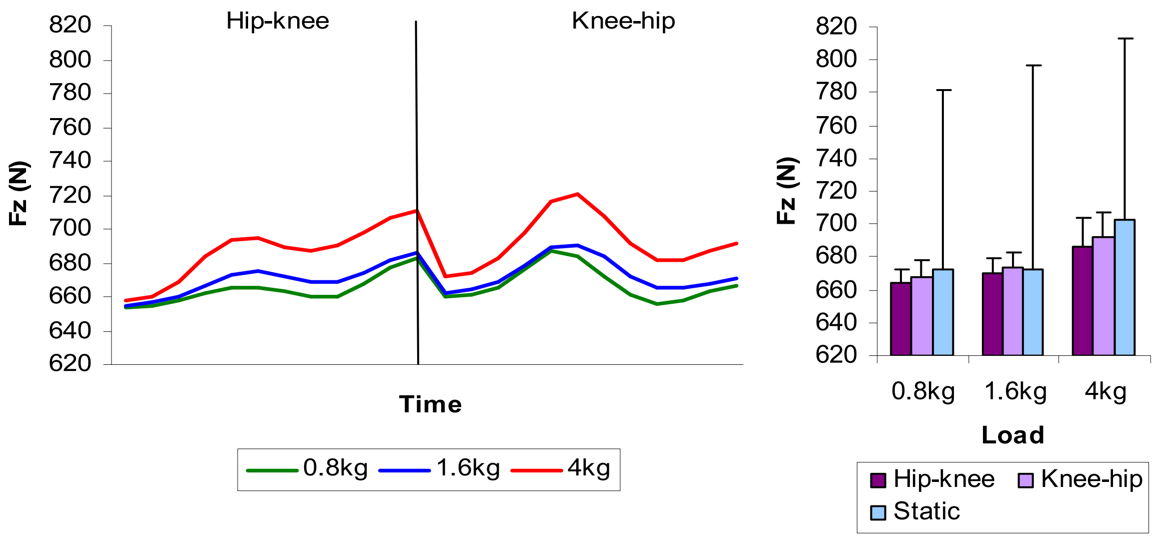
The 4kg loading condition resulted in greater sagittal displacement in comparison to the other loading conditions investigated and peaked at 5.5N in the hip-knee movement phase. In the 4kg loading condition, peaks in sagittal displacement occurred earlier in the hip knee movement phase and later in the knee hip movement phase compared to the other loading conditions (Figure 18A). Average sagittal ground reaction force was of a low magnitude and was negative in the knee-hip movement phase under all loading conditions. Average static sagittal ground reaction force differed slightly with increased load magnitude and peaked at 1.28N in the 4kg condition (Figure 18B).



(A) (B)

Figure 19: Pattern (A) and average data (B) lateral ground reaction forces under each loading condition during the "hip knee" task.

Lateral ground reaction forces were similar under all loading conditions in the "hip knee" task but differed more markedly in the hip-knee movement phase. Lateral displacement during the "hip knee" task was of a low magnitude during both movement phases with the exception of the initial phases of the knee-hip movement phase (Figure 19A). Average lateral ground reaction force was greatest in the hip-knee movement phase in all loading conditions with a value of -0.86N and did not differ markedly with increased external load. Average static lateral ground reaction force was lowest in the 4kg condition with an average value of -0.7N (Figure 19B).



(A) (B)

Figure 20: Pattern (A) and average (B) vertical ground reaction forces under each loading condition during the "hip knee" task.

Vertical ground reaction forces were greatest in the 4kg loading condition throughout task completion and peaked in the knee-hip movement phase at 720N. The pattern of the vertical ground reaction forces followed a similar trend under all loading conditions (Figure 20A). Average vertical ground reaction force was greater in the knee-hip movement phase than the hip-knee phase. Average static vertical ground reaction force was greater than that of the movement phases in the 0.8kg and 4kg condition with a value of 672N and 703N respectively (Figure 20B).

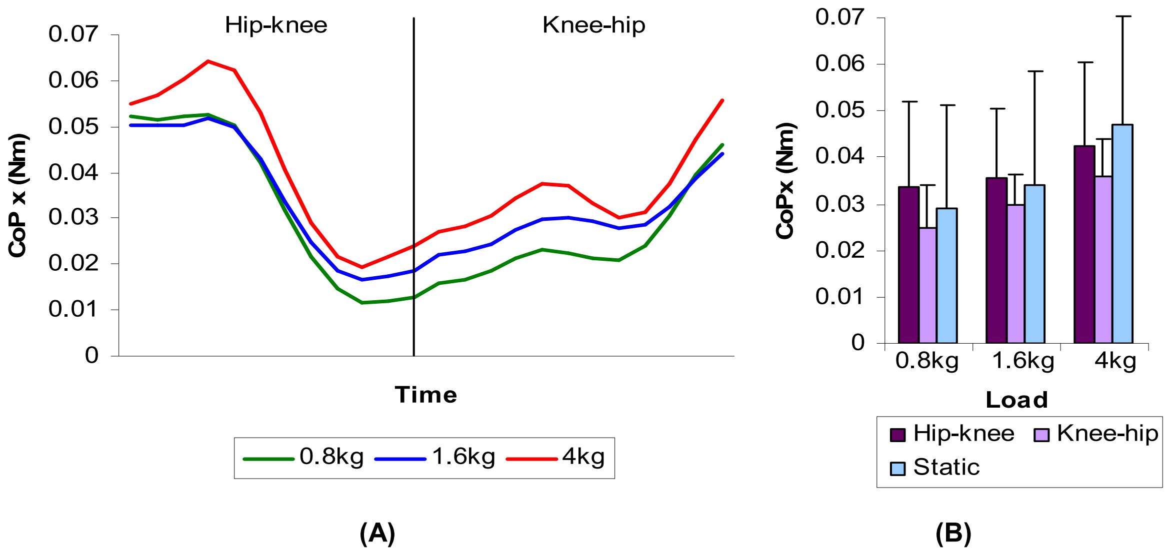


Figure 21: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the anterior/ posterior direction under each loading condition during the "hip knee" task.

Peaks and troughs in sagittal displacement of the COP occurred at the same point in time under all loading conditions but were more pronounced in the 4kg condition. The pattern of COP displacement remained positive throughout task completion (Figure 21A) and peaked in the hip-knee movement phase at 0.064Nm. Average COP displacement in the anterior/posterior direction was greatest in the hip-knee movement phase in comparison with the knee-hip movement phase under all loading conditions. Average static COP displacement was greatest in the 4kg condition with an average value of 0.04Nm, which was greater than that of both movement phases (Figure 21B).

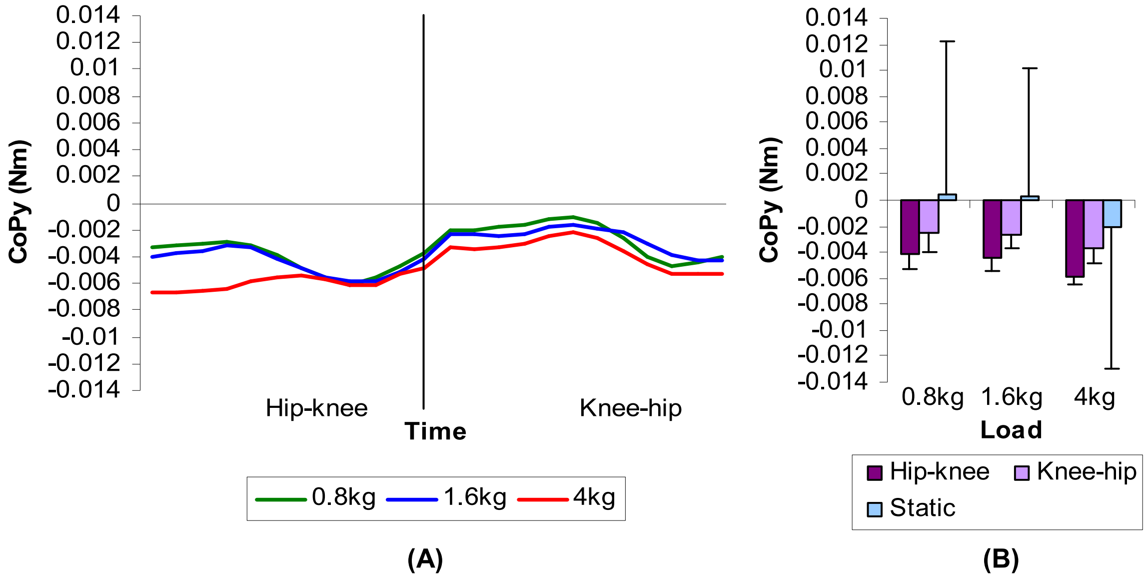


Figure 22: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the lateral direction under each loading condition during the "hip knee" task.

The lateral displacement of the COP remained negative throughout task completion under all loading conditions in the "hip knee" task (Figure 22A). Lateral displacement was of a low magnitude and followed a similar trend under each loading condition and was greatest in the hip-knee movement phase at 0.0067Nm. Average lateral COP displacement was negative in both movement phases of the "hip knee" task and increased with increased load magnitude (Figure 22B). Average static lateral displacement of the COP was positive in the 0.8kg and 1.6kg condition with a value of 0.0004Nm and 0.0002Nm respectively (Figure 22B).

LOAD COMPONENT OF POSTURAL CONTROL

Differential analysis was used to determine the "load component" of postural control. Data was manipulated such that the difference in muscle activity and ground reaction forces between load 2 (1.6kg) and load 1 (0.8kg) was multiplied by four. The difference in muscle activity and ground reaction forces between load 3 (4kg) and load 2 (1.6kg) was multiplied by three. If these differences were equal then the load component of postural control was linear in nature in nature. Design matrix illustrating differential analysis can be seen in Figure 5 (p.46).

MUSCLE ACTIVITY

Table III: Significant differences indicating the load component of postural control for all muscles and all tasks investigated.

TASK		LES (R)	LES (L)	TES (R)	TES (L)	EO (R)	EO (L)	RA (R)	RA (L)	IO (R)	IO (L)	LAT (R)	LAT (L)
"Hip knee"	1					↓						↓	↓
	2			↓	↑	↓	↓						↓
"Hip shoulder"	1	↓	↓	↓	↓	↓	↑		↓	↓	↑	↑	↑
	2	↑	↑	↑	↑	↓		↓	↑	↓	↑	↑	↑
"Hip twist"	1		↑	↓	↑						↑	↓	↑
	2	↓	↓	↑	↓		↓	↓	-	↓	↓	↓	↑
"Hip reach"	1	↓	↑	-	↓	↓	↓	↓	↓	↓	↓	↑	↓
	2	↓	↑	↓	↓	↑	↑	↓	↓	↓	↑	↓	↑

---- Condition significantly different (p<0.05) ----Condition*data point significantly different (p<0.05)

Where:

1 = first movement phase
2 = second movement phase
(R)= right
(L)= left
↓↑= direction of muscle activity from A-4B condition

LES= lumbar erector spinae
TES= thoracic erector spinae
EO= external oblique
RA= rectus abdominis
IO= internal oblique
LAT= latissimus dorsi

The "hip knee" task resulted in the fewest significant differences (p<0.05) in muscle activity pattern with 10 of the 19 significant results occurring as a result of the interaction between load and movement pattern (condition*data point). The presence of a significant effect (p<0.05) of increased load was not the same in the majority of right and left muscle pairs (Table III). This was particularly evident in the "hip twist" task which was expected due to the asymmetrical nature of the task. Increased load magnitude had a significant effect (p<0.05) on the muscle activity of left latissimus dorsi during both movement phases of all tasks investigated. Further, a significant interactional effect between condition and data point occurred in the left latissimus dorsi in all conditions. Increased load magnitude also had a significant effect on muscle activity in the "hip reach" task. Further, a significant interactional effect between condition and data point occurred in all muscles of the "hip reach" task, with the exception of the right and left lumbar erector spinae muscles during the second movement phase. As such, the "hip reach" task could be viewed as the most taxing of the tasks investigated.

GROUND REACTION FORCES

Table IV: Significant differences indicating the load component of postural control in all tasks investigated.

TASK		F_x		F_y		F_z		COP_x		COP_y	
"Hip knee"	1		x			#	x		x		
	2	#	x			#	x		x		
"Hip shoulder"	1	#	x			#	x		x	#	
	2		x				x	#	x	#	
"Hip twist"	1	#		#	x	#	x		x		x
	2	#		#		#	x	#	x	#	
"Hip reach"	1	#		#		#	x				
	2					#	x				

Where:
COP_x = Movement of the center of pressure in the sagittal plane
COP_y = Movement of the center of pressure in the lateral plane
F_x = Sagittal ground reaction force
F_y = Lateral ground reaction force
F_z= Vertical ground reaction force
1 = first movement phase
2 = second movement phase
# = condition significant ($p<0.05$)
x = condition*data point significant ($p<0.05$)

Altered data in the first movement phase of the "hip knee" task resulted in the significant effect of condition in only the vertical ground reaction forces (F_z). Pattern of ground reaction forces was significantly altered in both movement phases in the sagittal (F_x) and vertical (F_z) planes. The pattern of sagittal displacement of the COP was also significantly altered both movement phases of the "hip knee" task. Increased load magnitude resulted in significantly different ($p<0.05$) muscle activation and a significant interactional effect between condition and data point in vertical ground reaction forces during both movement phases of all tasks investigated, with the exception of the second movement phase of the "hip shoulder" task (Table IV). A significant interactional effect ($p<0.05$) between condition and data point occurred in the movement of COP in the sagittal plane (COP_x) in all tasks and movement phases with the exception of the "hip reach" task. The second movement phase of the "hip reach" task elicited the fewest significant differences and these occurred in the vertical ground reaction force (F_z). Increased load magnitude did not have a significant impact on lateral ground reaction force (F_y) and pattern of lateral ground reaction force in both movement phases of the "hip knee", "hip shoulder" and second movement phase of the "hip reach" task.

MUSCLE ACTIVITY

Table V: Significant differences indicating the movement component of postural control for all muscles and all tasks investigated.

TASK		LES (R)		LES (L)		TES (R)		TES (L)		EO (R)		EO (L)		RA (R)		RA (L)		IO (R)		IO (L)		LAT (R)		LAT (L)	
"Hip knee"	1		x		x		x		x		x	●	x		x		x	●	x		x		x		x
	2	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●		●	x	●	x	●	x	●	x
"Hip shoulder"	1	●	x	●	x	●	x	●	x		x		x			●	x	●	x	●	x		x	●	x
	2	●	x	●	x	●	x	●	x		x	●	x	●	x	●	x	●	x	●	x	●	x	●	x
"Hip twist"	1		x	●	x	●	x		x		x	●	x		x	●	x	●	x	●	x				x
	2	●	x	●	x	●	x		x		x	●	x		x	●	x		x	●	x		x	●	x
"Hip reach"	1	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x
	2	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x	●	x

● Load significant ($p<0.05$)
x Load*data point significant ($p<0.05$)

Where:

1 = first movement phase
2 = second movement phase
(R)= right
(L)= left

LES= lumbar erector spinae
TES= thoracic erector spinae
EO= external oblique
RA= rectus abdominis
IO= internal oblique
LAT= latissimus dorsi

Increased body movement rendered the fewest significant differences in the "hip knee" task. In the second movement phase of the "hip knee" task, all muscles investigated with the exception the left rectus abdominis muscle, elicited a significant interactional effect between load and data point, as well as a significant effect of load ($p<0.05$). This may, however be attributed to moving the load against gravity. In all tasks, a significant interactional effect ($p<0.05$) between load and data point occurred in the right and left lumbar erector spinae, right and left thoracic erector spinae, right and left external oblique, right and left internal oblique and the left latissimus dorsi muscle (Table V). The presence of significant differences in right and left muscle pairs differed during "hip twist" task. This was also evident in the other tasks but was not as pronounced. The "hip reach" task was characterised by a significant effect of load and an interactional effect between load and data point in all muscles investigated.

GROUND REACTION FORCES

Table VI: Significant differences in ground reaction forces indicating the movement component of postural control in all tasks investigated.

TASK		F_x		F_y		F_z		COP_x		COP_y	
"Hip knee"	1		x		x		x	#	x		x
	2		x				x	#	x		
"Hip shoulder"	1	#	x	#	x	#	x	#	x		x
	2		x				x	#	x		
"Hip twist"	1		x	#	x	#	x	#	x	#	x
	2	#	x	#	x	#	x	#	x	#	x
"Hip reach"	1					#	x	#	x		
	2	#	x	#	x	#	x	#	x	#	x

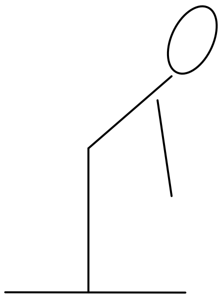
Where:
COP_x = Movement of the center of pressure in the sagittal plane
COP_y = Movement of the center of pressure in the lateral plane
F_x = Sagittal ground reaction force
F_y = Lateral ground reaction force
F_z= Vertical ground reaction force
1 = first movement phase
2 = second movement phase
# = condition significant ($p<0.05$)
x = condition*data point significant ($p<0.05$)

Pattern of movement of vertical ground reaction force (Fz) was significantly affected in altered data in all tasks investigated. Altered data revealed a significant effect of load and pattern of movement of the COP in the sagittal plane (COP_x) in both movement phases of all tasks investigated (Table VI). The first movement phase of the "hip shoulder" task rendered a significant effect of load and a significant interactional effect between load and data point in all ground reaction force variables with the exception of the effect of load on lateral COP movement (COP_y). The "hip twist" task resulted in the greatest number of significant differences as all ground reaction force variables elicited a significant effect of load as well as a significant interactional effect of load and data point, with the exception of sagittal ground reaction forces (F_x) in the first movement phase. In the second movement phase of the "hip reach" task, altered data resulted in a significant effect of load and a significant interactional effect between load and data point.

BODY DISCOMFORT

Body discomfort was assessed through the use of the Body Discomfort Map and Scale developed by Corlett and Bishop (1976). Subjects were required to identify and rate areas of discomfort after the completion of each condition. Areas of the body were then grouped into regions namely the upper limbs and neck, lower limbs, anterior trunk and posterior trunk. The intensity ratings of each condition were summed together to provide an overview of the condition experiencing the greatest levels of discomfort. The grouped areas of discomfort and ratings can be seen in Figure 23. A multivariate ANOVA and Tukey post-hoc test was performed and significant differences ($p<0.05$) were found between conditions and between loads. Total task discomfort for both static and dynamic conditions can be seen in Figure 24 (p.77) and Figure 25 (p.78).

“HIP KNEE”



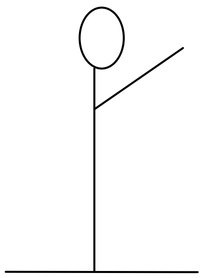
Posterior Trunk:
1: 28(2) 1. 53(2.5)
2: 32(2.6) 2. 62(2.48)
3: 44(2.4) 3. 100(3.03)

Lower Limbs:
1. 5(1.25) 1. 11(2.75)
2. 5(1.6) 2. 9(3)
3. -- 3. 13(1.857)

Upper Limbs and neck:
1. 3(3) 1. 3(3)
2. -- 2. 2(2)
3. 13(4.3) 3. 16(4)

Anterior Trunk:
1. -- 1. 3(3)
2. 3(1.5) 2. 12(3)
3. 2(2) 3. 26(3.25)

“HIP SHOULDER”

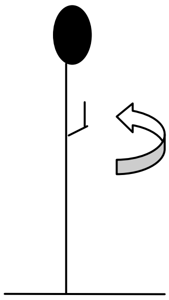


Posterior Trunk:
1: 4(1.3) 1. 6(1.5)
2: 12(1.7) 2. 17(2.43)
3: 25(3.125) 3. 29(2.63)

Upper Limbs and neck:
1. 15(1.66) 1. 11(2.2)
2. 49(2.7) 2. 62(2.58)
3. 102(3) 3. 96(2.82)

Anterior Trunk:
1. -- 1. --
2. -- 2. 2(2)
3. 5(2.5) 3. 5(1.66)

“HIP TWIST”



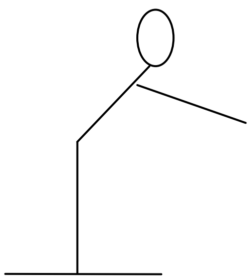
Posterior Trunk:
1: 12(1.7) 1. 15(2.5)
2: 13(1.8) 2. 21(2.63)
3: 30)2.7) 3. 38(2.92)

Lower Limbs:
1. -- 1. 1(1)
2. 2(2) 2. 2(2)
3. 1(1) 3. 3(1.5)

Upper Limbs and neck:
1. 1(1) 1. --
2. 8(2.6) 2. 6(2)
3. 34(2.26) 3. 23(2.875)

Anterior Trunk:
1. 4(1.3) 1. --
2. -- 2. 6(1.2)
3. 1(0.5) 3. 14(2)

“HIP REACH”



Posterior Trunk:
1: 36(2.4) 1. 32(2.28)
2: 50(2.5) 2. 55(2.62)
3: 99(3.3) 3. 66(3)

Lower Limbs:
1. 3(1.5) 1. 10(2)
2. 22(2.4) 2. 10(2.5)
3. 28(3.5) 3. 35(2.5)

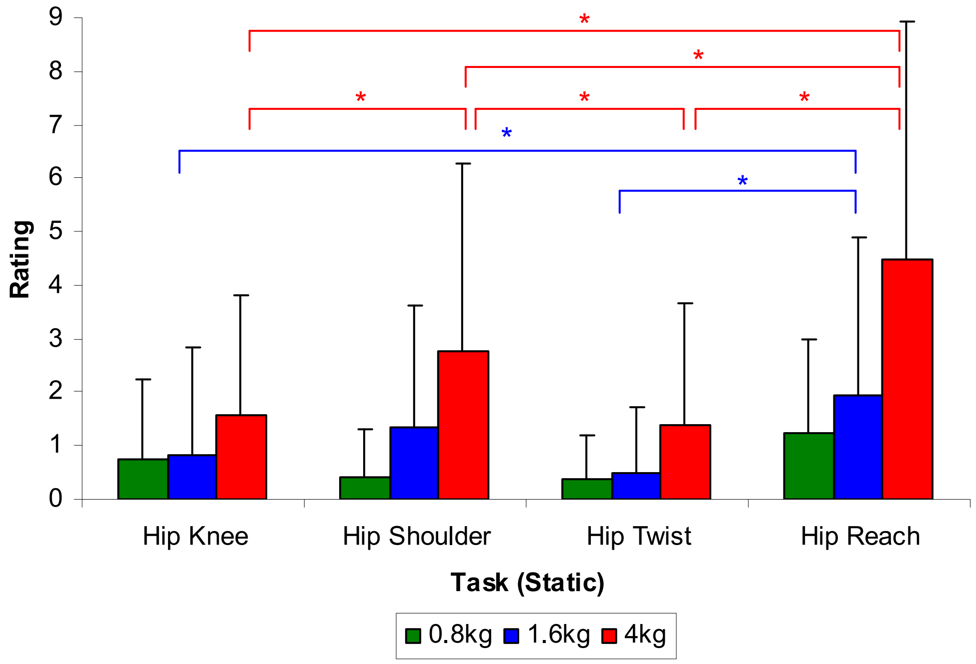
Upper Limbs and neck:
1. 5(1.6) 1. 10(2.5)
2. 4(2) 2. 7(2.33)
3. 51(3) 3. 42(3)

Anterior Trunk:
1. 4(2) 1. --
2. 9(1.8) 2. 1(1)
3. 33(3.6) 3. 15(2.1)

----- static condition ------- dynamic condition
1, 2, 3 refer to loading conditions (0.8kg, 1.6kg and 4kg)

Figure 23: Total and average rating of discomfort by body region during task completion (Total with average rating in brackets).

Discomfort ratings generally increased with increased load mass in both static and dynamic conditions of all tasks investigated. The lower limbs and anterior trunk in the “hip knee” task resulted in a decrease followed by an increase in total discomfort with increased load mass. Average rating did, however, increase in the anterior trunk in this condition. The same trend was found in the static “hip twist” task in the anterior trunk and lower limbs. During the “hip shoulder” task, subjects did not experience discomfort in the lower limbs.

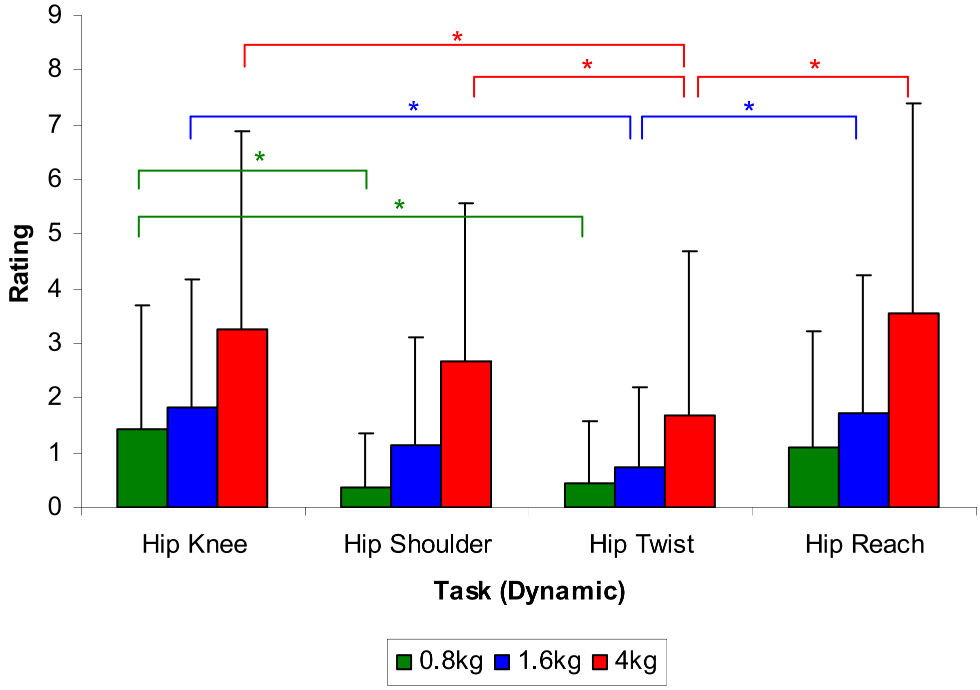


* Indicates significant difference ($p<0.05$) in the same loading condition

Figure 24: Total average static body discomfort rating with standard deviation bars according to task and load.

Average body discomfort associated with the static condition of each task increased under increased loading conditions for all tasks investigated. Average static body discomfort was greatest in the “hip reach” task under all loading conditions and peaked in the 4kg condition of the “hip reach” task at a rating of 4.47±4.4. Lowest static body discomfort occurred in the “hip twist” task in the 0.8kg condition and had an average rating of 0.37±0.8. No significant difference in body discomfort rating was found between tasks in the 0.8kg loading condition. In the 1.6kg condition, static discomfort ratings of the “hip reach” task was significantly greater than the “hip knee” and “hip twist” task. In the 4kg loading condition, discomfort associated with the “hip shoulder” task was significantly greater than that of the “hip knee” and “hip twist’ task.

Discomfort of the “hip reach” task in the 4kg loading condition was significantly greater than the discomfort experienced during the “hip twist”, “hip shoulder” and “hip knee” tasks.



* Indicates significant differences (p<0.05) in the same loading condition

Figure 25: Total average dynamic body discomfort rating with standard deviation bars according to task and load.

Increased load magnitude resulted in increased average body discomfort ratings for all tasks. Greatest average discomfort was experienced in the static “hip reach” task performed with the 4kg load and had an average rating of 4.48±4.5. Lowest average discomfort occurred in the 0.8kg condition of the “hip shoulder” task and had a rating of 0.35±0.9. Average body discomfort in the 0.8kg condition of the “hip knee” task was significantly greater than the discomfort of the “hip shoulder” and “hip twist” tasks under the same loading condition. In the 1.6kg condition, discomfort of the “hip reach” task was significantly greater than that of the “hip twist” task. Discomfort of the “hip twist” task was also significantly lower than that of the “hip knee” task under the same loading condition. In the 4kg condition, discomfort of the “hip twist” task was significantly lower than the “hip knee”, “hip shoulder” and “hip reach” tasks.

Discomfort between all conditions can be seen in Appendix D (p.227)

CHAPTER V
DISCUSSION

INTRODUCTION

Despite the great effort which has been placed into the reduction of low back disorder (LBD) risk and prevalence, the occurrence of LBDs and low back pain in still high (Davis and Marras, 2000). One reason for this may be due to the fact that developed guidelines do not account for the underlying mechanisms of postural control (Kollmitzer, 2002) which may place additional strain on the body in excess of those imposed by task demands. Consequently, this study sought to establish the impact of increased load and trunk movement on the muscle activation and ground reaction forces attributed to the maintenance of postural equilibrium.

LOAD COMPONENT OF POSTURAL CONTROL

In order to isolate the muscle activity attributed to increased load magnitude, data was manipulated such that the effect of load was negated and is described in Figure 5 (p.46). Through this analysis, the effect of increased load magnitude on muscle activity and ground reaction forces associated with dynamic tasks could be analysed. The presence of significant differences ($p<0.05$) in analysed trunk muscles and ground reaction forces indicate that the muscle activity and ground reaction force attributed to load does not increase in a linear manner with increased load magnitude. As a result, the proposed "load component" of postural control is not a linear function. Interactional effects were also present between increased load and the muscle activation and ground reaction force pattern.

COMMON TRENDS IN ELECTROMYOGRAPHY

Muscle activity attributed to increased load magnitude followed a similar trend across all tasks investigated and was not a linear function. This is indicative of additional strain on the body in addition to that imposed by task demands. Observed and expected increase in muscle activity attributed to increased load magnitude is illustrated in Figure 26.

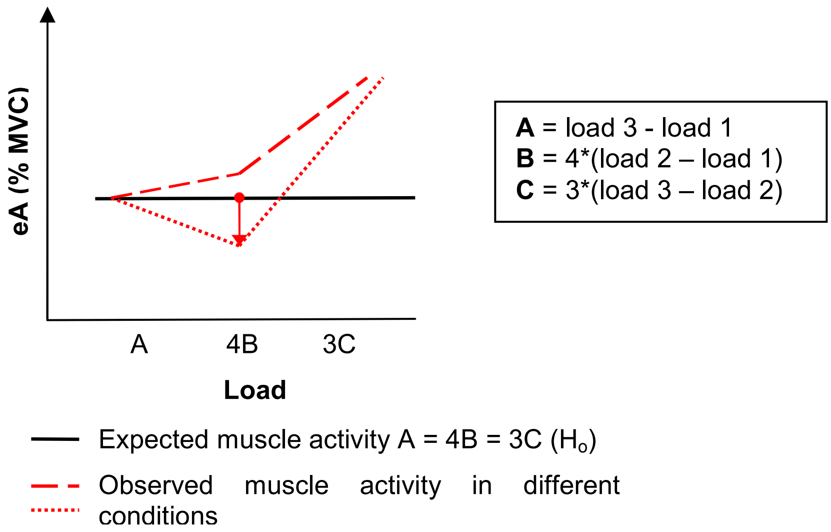


Figure 26: Expected and observed increases in muscle activity in manipulated data.

Muscle activity attributed to load resulted in both an increase and decrease in activity from the A to 4B condition (Figure 26). Decreases in altered data from the A to 4B condition is indicative of a degree of movement efficiency as muscle activity decreased with an increase in external load. Increased muscle activity from the A to 4B condition shows an increase in muscle activity attributed to load magnitude, over and above the effects of external load. Muscle activity associated with the 3C condition was greater than that of the A condition in all muscles which were significantly affected in altered data (Figure 26). Therefore, muscle activity attributed to load increased more substantially when manipulating greater external loads.

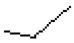
COMMON TRENDS IN GROUND REACTION FORCES

The pattern of vertical ground reaction force attributed to load was similar in the "hip shoulder", "hip twist" and "hip reach" tasks (Appendix C Figure 77:182; Figure 82:185; Figure 87:187) but differed in the "hip knee" task (Figure 20). This could be attributed to the nature of the tasks investigated as the "hip knee" task was the only task which required the movement of loads close to floor level and consequently required a large degree of trunk flexion.

Increased load magnitude resulted in greater displacement in the sagittal plane (F_x) in all tasks investigated (Figure 18:67; Figure 75:181; Figure 80:184 and Figure

85:186). Therefore, increased load magnitude caused greater forward and backward motion of sagittal ground reaction forces with increased load magnitude. This could be attributed to shifting the mass of the body further forward and backward in order to counteract the increased external load which was positioned in front of the body.

Table VII: Trend of ground reaction forces attributed to increased load magnitude of all tasks investigated.

TASK		F_x	F_y	F_z	CoP_x	CoP_y
"Hip knee"	1					
	2					
"Hip shoulder"	1					
	2					
"Hip twist"	1					
	2					
"Hip reach"	1					
	2					

Where: F_x = Sagittal ground reaction force
F_y = Lateral ground reaction force
F_z= Vertical ground reaction force
COP_x = Movement of the center of pressure in the sagittal plane
COP_y = Movement of the center of pressure in the lateral plane

1 = first movement phase
2 = second movement phase

Sagittal ground reaction force attributed to increased external load (Table IV:72) was significantly altered in the second movement phase of the "hip knee" task, the first movement phase of the "hip shoulder" and "hip reach" tasks and in both movement phases of the "hip twist" task. The trend of sagittal ground reaction force attributed to increased external load increased from the A to 4B condition and subsequently decreased in the 3C condition to a lower level than the A condition (Table VII). This trend occurred in all the above tasks with the exception of the first movement phase of the "hip twist" task. Therefore, sagittal ground reaction force increased when manipulating light external loads (0.8kg, 1.6kg) and decreased under heavier loading indicating lower sagittal displacement in such conditions. The first movement phase of the "hip twist" task elicited the opposite trend as sagittal movement attributed to external load was greater in the 4C condition than in the A condition and decreased from the A to 4B condition (Table VII). Thus, when twisting towards the dominant side, anterior/posterior movement attributed to load decreased under external loads of 0.8kg and 1.6kg but increased under heavier loading conditions.

The absence of a significant effect of load on lateral ground reaction force in the “hip knee”, “hip shoulder” and “hip reach” tasks was expected as the tasks were symmetrical in nature. Increased load magnitude did, however, have a significant effect on manipulated lateral ground reaction force in the first movement phase of the “hip reach” task. Thus, manipulating loads from hip height to a condition of extended horizontal reach resulted in a decrease in lateral movement attributed to load under light external loads. Heavier loads did, however, result in greater lateral ground reaction force attributed to load (Table VII:81). The presence of a significant effect of load on lateral ground reaction force in both movement phases of the “hip twist” task was expected as the task was asymmetrical in nature. Subjects twisted to the right and shifted body mass onto the left leg as indicated by negative F_x values in the hip-twist movement phase (Appendix C Figure 80:184). Weight was shifted to the right when subjects returned to an upright posture. This effect was significantly different under increased loading conditions (Table IV:72).

Increased load magnitude had a significant effect on vertical ground reaction forces in both movement phases of all tasks, with the exception of the second movement phase of the “hip shoulder” task. Additionally, the trend of vertical ground reaction force attributed to load was the same in all tasks given that the 3C condition was associated with greater vertical ground reaction force in comparison with the A condition. Vertical ground reaction force attributed to load magnitude did, however, decrease from the A to 4B condition in the second movement phase of the “hip knee” task and during both movement phases of the “hip twist” task (Table VII:81). Therefore, even when the effect of load was negated, vertical ground reaction forces were significantly affected by increased load magnitude. The sharp decrease in vertical ground reaction force at the onset of the second movement phase in the “hip knee” (Figure 20:68) “hip shoulder” (Appendix C Figure 77:182) “hip twist” (Appendix C Figure 82:185) and “hip reach” tasks (Appendix C Figure 87:187) may have been due to subjects leaning on the box at the end of the first movement phase and subsequently pushing off the box to initiate movement.

Movement of the centre of pressure in the horizontal plane (COP_x) provides valid information regarding postural stability (Karlsson and Frykberg, 2000). Increased load magnitude resulted in greater postural instability, indicated by greater movement of the COP in the horizontal plane in all tasks investigated (Figure 21:69; Figure 78:183; Figure 83:185 and Figure 88:188). Movement of the COP in the lateral plane (COP_y) remained negative during the “hip knee” (Figure 22:70) and “hip shoulder” tasks (Appendix C Figure 79:183). As a result, subjects shifted their weight to the left during task completion. As both of these tasks were symmetrical, lateral movement of the COP may have been due to the stance of subjects on the force platform as the feet may not have been exactly placed.

“LOAD COMPONENT” IN ALL TASKS INVESTIGATED

“Hip knee” task

Muscle activity

The “hip knee” task rendered the fewest significant differences in muscle activation during both movement phases (Table III:71). The first movement phase of the “hip knee” task required the load to be lowered from hip height to knee height. In this movement phase, significant differences were found in manipulated data in the right external oblique and the right and left latissimus dorsi muscles (Table III: 71). The second movement phase of the “hip-knee” task (lifting the load from knee height to hip height) elicited a significant effect of load in the right and left thoracic erector spinae, the right and left external oblique and the left latissimus dorsi muscles (Table III: 71). Consequently, increased external load resulted in a non-linear increase in muscle activity in the above muscles. This is indicative of additional strain on the musculoskeletal system in addition to that of task demands with increased load.

All muscles which resulted in a significant effect of increased load in the “hip knee” task, with the exception of the left thoracic erector spinae in the second movement phase, showed a decrease in muscle activity from the A to 4B condition (Table III; Figure 26:80). This decrease in muscle activity may be indicative of movement efficiency when manipulating light external loads as a lower level of muscle activity was required for task completion.

Davis *et al.* (1998) state that the muscle activity associated with lowering motions is lower than lifting exertions. The absence of a significant effect of load in nine of the 12 muscles in the first movement phase (Table III: 71), together with the lower average muscle activation levels (Figure 12 to Figure 17) during the hip-knee movement phase supports this literature. Additionally, back extensor activity has been shown to be lower during lowering tasks in comparison to lifting tasks (Henriksson *et al.,* 1972; Kumar and Davis, 1983; Marras and Mirka, 1989; de Looze *et al.,* 1993; Cresswell and Thorstensson, 1994). The lower muscle activity of the right and left lumbar and thoracic erector spinae muscles during the hip-knee movement phase (Figure 12:62 and Figure 13:63) as well as the absence of significant differences during the first movement phase in the same muscles (Table III: 71) is in support of this literature. Increased load magnitude affected muscle activity differently during both movement phases of the “hip knee” task. This can be attributed to moving the load in the same direction as gravity during the first movement phase (hip-knee) and against gravity in the second movement phase (knee-hip). This may also account for the decreased level of muscle activity during the hip-knee movement phase.

The absence of a significant effect of load in manipulated data in the right and left lumbar erector spinae muscle during both movement phases may be a result of the flexion relaxation phenomenon. The presence of the flexion relaxation phenomenon and subsequent myoelectrical silence (Kippers and Parker, 1984; Christopher *et al.,* 2005) when the load was positioned at knee height may have resulted in lower muscle activation as body weight may have been supported by the intervertebral discs and spinal ligaments (McGill and Kippers, 1994). In turn this would reduce the effect of increased load on lumbar muscle activity. In addition, subjects lifting technique may have contributed to decreased lumbar muscle activity when the load was positioned at knee height as some subjects maintained a straight back during task completion. This would result in greater muscle activity levels when performing the “hip knee” task under different loading conditions due to the maintenance lumbar lordosis. Other subjects rounded the back and allowed the spine to curve, resulting in decreased lumbar erector spinae activation.

The “hip knee” task was symmetrical in nature and the external load was placed directly in front of the subject. Therefore, similar activation of the right and left lumbar and thoracic erector spinae muscles was expected. Differences in activation of the right and left side of the body within specific muscles could be attributed to differences in muscle strength between the right and left sides of the body. All subjects were right hand dominant; as such the right hand side of the body could be expected to be stronger than the left. The lower muscle strength of the left hand side of the body would require a greater level of muscle activity to maintain force production in comparison with the right hand side muscles. Although the right and left rectus abdominis in the first movement phase, and right and left internal oblique in the second movement phase were not significantly affected by increased load magnitude, muscle activity of the left side of both muscles was significantly greater ($p<0.05$) than the right. Additionally, the muscle activity pattern of the rectus abdominis in the second movement phase was significantly different between right and left rectus abdominis muscles.

Ground reaction forces

During the first movement phase of the “hip knee” task, only the vertical ground reaction force attributed to increased load were significantly affected in altered data. As such, vertical ground reaction force attributed to increased load magnitude did not increase in a linear manner during the “hip knee” task. In the second movement phase of the “hip knee” task, sagittal ground reaction force (F_x) and vertical ground reaction force (F_z) were significantly altered in manipulated data (Table IV:72). Sagittal ground reaction force increased and vertical ground reaction force decreased from the A to 4B condition (Table VII:81). Therefore, small increases in external load resulted in increased sagittal movement which was accompanied by a decrease in vertical ground reaction force. From the 4B to 3C condition, sagittal movement decreased and vertical ground reaction force increased. Thus, sagittal ground reaction force attributed to load decreased under greater loading conditions, which was accompanied by an increase in vertical ground reaction force. Therefore, manipulating loads from knee height to hip height under higher loading conditions, ground reaction force attributed to increased load resulted in lower sagittal movement and greater vertical ground reaction force.

The hip-knee movement phase was associated with greater postural instability, indicated by greater variability in sagittal COP movement with increased load magnitude, in comparison with the knee-hip movement phase (Figure 21:69). This data supports that of Lee *et al.* (2008) who found that trunk flexion exertions were associated with poorer levels of postural control when compared to trunk extension exertions.

"Hip shoulder" task

Muscle activity

During the "hip shoulder" task, all muscles, with the exception of the right rectus abdominis in the first movement phase and the left external oblique in the second movement phase were characterised by a significant effect of load on muscle activation in manipulated data (Table III:71). Consequently, increased load magnitude resulted in a non-linear increase in muscle activity attributed to load when lifting investigated load magnitudes from hip to shoulder height (Table III: 71).

During the first movement phase of the "hip shoulder" task, the right and left lumbar and thoracic erector spinae muscles were characterised by a decrease in muscle activity between the A and 4B condition (Figure 26:80). The right and left external oblique muscles differed in response as the right external oblique showed a decrease in muscle activity from condition A to 4B where the left external oblique showed an increase in muscle activity between the two manipulated data conditions. The right and left internal oblique followed the same trend as the external oblique muscle between the two conditions. These findings indicate that movement efficiency and associated decrease in muscle activity occurred in the right and left paraspinal muscles as well as the right external oblique, right rectus abdominis and right internal oblique muscles when manipulating light external loads from hip height to shoulder height. Muscle activity of the right and left latissimus dorsi muscle increased from the A to 4B condition (Table III: 71) which can be attributed to the function of the latissimus dorsi muscle, which extends abducts and medially rotates the arm at the shoulder joint (Tortora and Grabowski, 2003).

The right and left rectus abdominis muscle responded differently during the first movement phase of the “hip shoulder” task in manipulated data (Table III:71). The presence of a significant difference ($p<0.05$) in the left rectus abdominis may be attributed to strength differences between the right and left sides, as the left rectus abdominis would require greater activation to counteract the same external load. During the “hip shoulder” task, the rectus abdominis was not involved in the primary motor task as the trunk remained upright during task completion. In this task the rectus abdominis muscles may have been activated consistently throughout task completion, contributing to intra abdominal pressure, thereby aiding in the maintenance of spinal stability (Cholewicki and McGill, 1996; Cholewicki *et al*., 1997) and counteracting the effect of the external load placed in front of the body. This is illustrated in Appendix C (Figure 45:168 and Figure 46:169) which shows approximately a 1% MVC increase in muscle activity with increased load magnitude.

In the second movement phase of the “hip shoulder” task, muscle activity increased from the A to 4B condition in the thoracic and lumbar erector spinae muscles, right internal oblique, left rectus abdominis as well as the right and left latissimus dorsi (Table III; Figure 26:80). Therefore, even when the effect of load was negated, muscle activity increased with increased load magnitude when lowering loads from shoulder height to hip height. As such, movement efficiency and subsequent decrease in muscle activity does not occur in the thoracic and lumbar erector spinae and the left rectus abdominis muscle when manipulating relatively light external loads. This can be attributed to a greater need to control and decelerate the external load while lowering the load to hip height. The “hip shoulder” task was characterised by greater mechanical disadvantage at shoulder height as the arms were extended at shoulder level. Therefore, increased activity of the lumbar and thoracic erector spinae muscles was necessary to counteract the effect of the load when the arms were extended. The arms were the primary movers in the “hip shoulder” task and as such, increased activity of the latissimus dorsi can be attributed to increased need to control and decelerate the load against gravity. In contrast to the above muscles, muscle activity decreased from the A to 4B condition in the right external oblique, right rectus abdominis muscle and the right internal oblique muscle. This indicates a degree of movement efficiency when manipulating light external loads in these muscles.

Ground reaction forces

In the first movement phase of the "hip shoulder" task, sagittal and vertical ground reaction force as well as lateral movement of the COP (COP_y) was significantly altered in manipulated data (Table IV:72). This indicates that when moving loads from hip height to shoulder height, anterior/posterior and vertical ground reaction force attributed to load do not increase in a linear manner with increased load magnitude. Trend of sagittal ground reaction force attributed to increased load has been discussed previously (Table VII:81). Vertical ground reaction force and lateral movement of the COP followed a similar trend with increased load magnitude as ground reaction forces associated with the 3C condition were greater than that of the A condition. Lateral COP movement did however decrease from the A to 4B condition where vertical ground reaction force increased (Table VII:81). Significantly altered sagittal ground reaction force in the first movement phase was expected due to the nature of the task as greater sagittal trunk motion occurred to compensate for the load and to reduce the effort required in the upper extremities. Thus, manipulating loads from hip height to shoulder height when the effect of load was negated, resulted in increased vertical ground reaction force and lateral COP movement at a higher load magnitude. A degree of efficiency occurred between the A and 4B condition in lateral COP movement as COP movement decreased with increased external load.

In the second movement phase of the "hip shoulder" task, movement of the centre of pressure in the sagittal and lateral planes was significantly altered in manipulated data (Table IV:72). This indicates a non-linear increase in COP movement with increased load magnitude. Trend of COP movement attributed to load was similar in both the sagittal and lateral planes and increased with increased load magnitude with the 3C condition eliciting a greater response than the A condition (Table VII:81). Lateral COP movement did, however, decrease from the A to 4B condition indicating a reduction in lateral movement when manipulating light external loads (Table

VII:81). Significant sagittal COP movement was expected due to the nature of the task as previously discussed. It is, however, interesting to note that increased load resulted in a non linear increase in lateral COP movement. This indicates that greater sideward movement of the COP with increased load magnitude (Figure 79:183) and this increase is not a linear function.

"Hip twist" task

Muscle activity

In the first movement phase of the "hip twist" task, muscle activity attributed to load increased from the A to 4B condition in the left lumbar erector spinae, left thoracic erector spinae, left internal oblique and left latissimus dorsi muscles (Table III: 71). In contrast, muscle activity attributed to load, decreased from the A to 4B condition in the right thoracic erector spinae and right latissimus dorsi muscles. Neither the right nor left rectus abdominis muscle was significantly influenced in altered data during the first movement phase, but muscle activity attributed to load of both the right and left rectus abdominis was significantly impacted by increased load magnitude in the second movement phase.

The absence of a significant effect ($p<0.05$) of increased load magnitude in the right lumbar erector spinae could be attributed to the nature of the task. During the first movement phase, the right lumbar erector spinae muscle may have been recruited to a greater degree than the left lumbar erector spinae under all loading conditions, in order to twist the trunk to the right for load placement. This may result in the absence of significant differences due to constant muscle activation throughout task completion. The muscle activation pattern displayed in Appendix C (Figure 51:171 and Figure 52:171) supports this notion as increased load magnitude had a more pronounced effect on the left lumbar erector spinae muscle in comparison with the right lumbar erector spinae.

The rectus abdominis muscles were not extensively activated during the "hip twist" task as activity level ranged from 1.3-2.4% MVC in the first movement phase (hip-twist) and 1.1-1.7% MVC in the second (twist-hip) movement phase (Appendix C Figure 57:174). In the second movement phase, muscle activation of the right and left rectus abdominis was significantly affected in the altered data indicating a non-linear increase in muscle activity with increases in load.

During the second movement phase of the “hip twist” task, muscle activity decreased from the A to 4B condition in the right and left lumbar erector spinae, left thoracic erector spinae, left external oblique, right rectus abdominis, right and left internal oblique and right latissimus dorsi muscles. These data are indicative of a degree of movement efficiency when manipulating light external loads towards the non-dominant side. Activity of the left rectus abdominis muscle increased as a linear function from the A to 4B condition. Muscle activity attributed to load did, however, increase under heavier loading conditions. According to this data, when performing twisting tasks towards the dominant side, increased load magnitude does not place additional strain on the left rectus abdominis. Additional loads in excess of 1.6kg did, however, result in additional strain. In the second movement phase, muscle activity attributed to load increased from the A to 4B condition in the right thoracic erector spinae and left latissimus dorsi muscles. Increased muscle activity of the left latissimus dorsi muscle with increased load magnitude could be due to the role of the latissimus dorsi muscle in extending the arm at the shoulder joint (Tortora and Grabowski, 2003). At the beginning of the second movement phase, the left arm was extended to a greater degree than the right. The load was lifted in this posture and therefore, due to increased mechanical disadvantage as a result of the extended arm, the left latissimus dorsi muscle may produce greater muscle force in order to counteract the load.

During asymmetrical lifting tasks, loads are supported by the oblique muscles rather than the erector spinae muscles (Marras and Mirka, 1992). This investigation supports this, as the right and left external and internal oblique were activated differently during the both movement phases of the “hip twist” task. Although the right external oblique muscle was not significantly affected in the altered data, muscle activity of the right external oblique increased towards load placement in the hip-twist phase of movement (Appendix C Figure 55:173). This can be attributed to the function of the oblique muscles which rotate the vertebral column when acting unilaterally (Tortora and Grabowski, 2003). The activity of the left external oblique in the first movement phase increased from the A to 4B condition (Figure 26:80). The left external oblique was not the prime mover in the first movement phase of the "hip twist" task and therefore increases in muscle activation under relatively light loads could be a result of increased muscle activity to counteract the load. This finding could also be attributed to movement characteristics during task completion. Although subjects were instructed not to twist the hips during this task, subjects may have rotated the hips to aid in task completion, thereby reducing the activity of the right external oblique.

Ground reaction forces

Sagittal, lateral and vertical ground reaction forces were significantly altered with increased load magnitude in manipulated data during both movement phases of the "hip twist" task (Table IV:72). The trend of sagittal and lateral ground reaction force differed during both movement phases of the "hip twist" task. The first movement phase was characterized by greater sagittal ground reaction force in the 3C condition in comparison to the A condition. The opposite occurred in the second movement phase (Table VII: 81). The presence of a significant impact of load on sagittal ground reaction force in the "hip twist" task was unexpected as the task was symmetrical in nature and required minimal flexion and extension in the sagittal plane as the working height remained constant. However, significantly altered lateral ground reaction force with increased load magnitude in the "hip twist" task was expected due to the nature of the task. Additionally, greater external load was expected to increase lateral movement. Trend of lateral ground reaction force was different between the movement phases as the first movement phase resulted in lower sagittal ground reaction force in the 3C condition in comparison with the A condition. The opposite occurred in the second movement phase of the "hip twist" task (Table VII:81). From this data it can be concluded that twists towards the dominant side (first movement phase) resulted in greater sagittal ground reaction force and lower lateral movement under heavier loads. Twisting from the dominant side towards the midline was accompanied by lower sagittal ground reaction forces and greater lateral movement with increased load magnitude. Sagittal and lateral COP movement also did not increase in a linear manner in the second movement phase of the "hip twist" task (Table IV: 72). Sagittal COP movement associated with increased load increased under light external loads but decreased under greater loading conditions (Table

VII:81). Lateral COP movement associated with increased load magnitude decreased from the A to 4B condition and increased to the 3C condition (Table VII:81).

Therefore, when twisting towards the non-dominant side, increased load magnitude resulted in decreased sagittal and increased lateral COP movement. The non-linear increases in ground reaction force and COP movement with increased load magnitude in the "hip twist' task supports literature which states that asymmetrical lifting is associated with the development of more complex trunk motion patterns (Ferguson and Marras, 1992; Ferguson *et al.,* 1992; Allread *et al.,* 1996) and a reduction in trunk strength (Marras and Mirka, 1989; Ferguson *et al.,* 1992).

"Hip reach" task

Muscle activity

Muscle activity attributed to load was significantly affected by increased load magnitude in manipulated data in all muscles during the first movement phase of the "hip reach" task (Table III: 71). This data indicates that muscle activity did not increase in a linear manner with increased external load, even when the effect of the load was negated.

In the first movement phase of the "hip reach" task, muscle activity increased from the A to 4B condition in the left lumbar erector spinae and right latissimus dorsi muscles (Figure 26:80). The increase in activity of the left lumbar erector spinae could indicate strength differences between the right and left sides as the lumbar erector spinae muscles would be activated to counteract the flexion movement during the first movement phase. Although the right and left lumbar erector spinae were significantly affected by increased load magnitude in the first movement phase of the "hip reach" task, muscle activity was significantly greater ($p<0.05$) in the left lumbar erector spinae than the right lumbar erector spinae. Muscle activity of the right latissimus dorsi increased from the A to 3C condition (Table IV:72) which could be attributed to uneven box placement. Subjects were right hand dominant and subsequently tended to place the box unevenly with the left side placed before the right. This could result in increased muscle activation of the right side of the body, resulting in an increase in muscle activity between the A and 4B condition.

In the first movement phase of the “hip reach” task, the right and left muscles of the external oblique, rectus abdominis and internal oblique responded in the same manner eliciting a decrease in muscle activity between the A and 4B condition (Table III: 71). As a result, these muscles experienced a degree of movement efficiency during the first movement phase of the “hip reach” task between the 0.8kg and 1.6kg loading condition. In the first phase of movement, the control of the external load and the maintenance of postural equilibrium may have shifted to the muscles of the lower extremity, reducing the strain on the external oblique, rectus abdominis and internal oblique muscles. During task performance, subjects bent the knees slightly in order to avoid falling. Therefore, the muscles of the lower limb may have contributed to task completion, resulting in a decrease in muscle activity from the A to 4B condition of the above muscle pairs.

The second movement phase of the “hip reach” task required subjects to return the load to the hips. In this movement phase, muscle activity decreased from the A to 4B condition in the right lumbar erector spinae, right and left thoracic erector spinae, right and left rectus abdominis, right internal oblique and right latissimus dorsi muscles (Table III: 71). Conversely, muscle activity increased in the left internal oblique and latissimus dorsi muscles. This may indicate increased muscle recruitment of the left muscles in order to produce the desired force to control load movement. The decrease in muscle activity of the right lumbar erector spinae and right and left thoracic erector spinae from the A to 4B condition could be due to subjects using box placement at the reaching phase to push themselves towards an upright position. This would reduce the contribution of the erector spinae muscles to the extension of the vertebral column when manipulating light external loads. The decrease in muscle activation between the A and 4B condition of the rectus abdominis muscles can be attributed to the role of the rectus abdominis muscles which flex the vertebral column (Tortora and Grawbowski, 2003). As such, decreased muscle activity during the second movement phase is not unexpected. Muscle activity of the right thoracic erector spinae increased in a linear manner between the A and 4B condition during the second movement phase of the “hip reach” task. Therefore, the maintenance of postural equilibrium when manipulating light external loads with increased horizontal reach does not require additional muscular effort in the right thoracic erector spinae muscle.

Overall the "hip reach" task was the most affected by increased load magnitude as muscle activity of all muscles was significantly affected ($p<0.05$) by increased in load magnitude even when the effect of the load was negated. This was, however, expected as increases in load magnitude and performing tasks with increased horizontal reach, places the body at an increased mechanical disadvantage (Ferguson *et al.,* 1992; Jorgensen *et al.,* 2003). With increased horizontal reach, the torso muscles increase force production to counteract the increased external force generated by increased moment arm. This effect is magnified with increases in load magnitude as a result of the generation of greater external force on the body (Jorgensen *et al.,* 2003).

IMPACT OF LOAD ON MUSCLE ACTIVITY PATTERN

In the first movement phase of the "hip knee" task, a significant interactional effect between the altered data and the muscle activation pattern was found in the right and left thoracic erector spinae, the right internal oblique and right and left latissimus dorsi. Manipulating the data such that the effect of load was negated, resulted in significantly different muscle recruitment patterns in these muscles. The pattern of muscle activation of the right and left latissimus dorsi muscles was also significantly different ($p<0.05$) to each other. In the second movement phase, pattern of muscle activity was significantly affected by increased load magnitude in the right and left lumbar erector spinae, right external oblique, right rectus abdominis and the left latissimus dorsi muscles (Table III: 71). The first movement phase was associated with a smaller degree of variability in muscle activation pattern in comparison with the second movement phase as fewer muscles rendered significantly different ($p<0.05$) activation patterns during the first movement phase.

In both movement phases of the "hip shoulder" task, manipulated data significantly affected the pattern of muscle activation in the lumbar and thoracic erector spinae, the external oblique, internal oblique and latissimus dorsi muscle pairs (Table III: 71). During the second movement phase, muscle activity pattern of the left rectus abdominis was not significantly altered once data was manipulated. Additionally, muscle activation pattern of the right rectus abdominis was significantly different to the left rectus abdominis. The absence of a significant interactional effect of muscle recruitment pattern and load magnitude of the left rectus abdominis muscle may be due to constant activation with a small variability in muscle recruitment pattern as discussed previously. Peaks in muscle activity of the right rectus abdominis in the hip-shoulder movement phase can be attributed to control of trunk movement as subjects tended to lean backwards to aid in lifting the load to shoulder height (Appendix C Figure 45:168). Increased rectus abdominis activity at this point serves to counteract spine extension. In muscle activation pattern, the right rectus abdominis elicited a different muscle activity pattern than the left rectus abdominis. Subjects may have favoured the right rectus abdominis to oppose the backward lean of the body resulting in greater variation of muscle activity pattern in the right rectus abdominis compared to the left rectus abdominis muscle. Right and left rectus abdominis did however, have similar levels of muscle activation during the “hip shoulder” task. Muscle activity pattern peaked following load placement in all investigated muscles during the “hip shoulder” task (Appendix C Figure 39:166 to Figure 50:170) which could be attributed to the initial muscle force required to pick up the load at shoulder height.

With regard to the “hip twist” task, a significant interactional effect between load and muscle activity pattern occurred in both movement phases in the right and left lumbar erector spinae, right and left thoracic erector spinae, left external oblique, left internal oblique and left latissimus dorsi muscles (Table III:71). This indicates greater variability in muscle activity attributed to load when performing twisting tasks. The interactional effect of load and muscle recruitment pattern in the left muscles may be a result of strength differences as muscle activation pattern would differ more markedly in muscles with lower strength capabilities. Slight changes in muscle activity pattern of the above muscles are illustrated in Appendix C (p.163). Pattern of muscle activity was influenced to a greater extent in the second movement phase of the “hip twist” task as more muscles displayed a significant interactional effect between load and pattern of muscle activity in this movement phase (Table III: 71). This may be a result of task characteristics as the trunk and load were twisted towards the non-dominant side. The findings of this investigation are in accordance with that of Ferguson and Marras (1992), Ferguson *et al.* (1992) and Allread *et al.* (1996) who proposed that asymmetrical lifting is associated with the development of more complex trunk motion patterns. In this investigation, this is illustrated by a greater complexity in muscle recruitment patterns during the “hip twist” task.

Increased load magnitude had the greatest effect on muscle activation pattern during the “hip reach” task as a significant interactional effect between load and data point occurred in all muscles, with the exception of the right and left lumbar erector spinae muscles in the second movement phase. This absence of a significant interactional effect can be attributed to the consistent trend in muscle activation pattern throughout the reach-hip movement phase in the lumbar erector spinae muscles, an example of which can be seen in Appendix C (Figure 63:177). Muscles were constantly activated to counteract the effect of increased mechanical disadvantage as a result of body posture. Although muscle activity pattern was not significantly altered in the right and left lumbar erector spinae, slight changes in muscle activity pattern of the right thoracic erector spinae with increased load magnitude were observed and are shown in (Appendix C Figure 65:178). Further, muscle activation pattern of the right and left lumbar and thoracic erector spinae muscles followed a similar trend with increased load magnitude (Appendix C Figure 63:177 to Figure 66:178). In the second movement phase of the “hip reach” task, muscle recruitment pattern in altered data of the right and left latissimus dorsi were significantly different to each other. This can also be attributed to strength differences between the right and left muscles.

Trunk muscle recruitment patterns have been shown to differ with increased external load (Fathallah *et al.,* 1998; de Looze *et al.,* 1999; Davis and Marras, 2000). The results of this investigation supports this literature, specifically when manipulating loads between hip and shoulder height and in tasks requiring extended horizontal reach as significant interactional effects between load and muscle activity were found extensively in these tasks (“Hip shoulder” and “hip reach” tasks).

Butler *et al.* (2008) investigated the pattern of trunk muscle activity while performing an upright standing normal reach and maximal reach task. During task completion, two antagonistic recruitment strategies were identified. Firstly, synergistic muscle activity occurred among all abdominal sites (Butler *et al.,* 2008). Data from this investigation supports the findings of these authors as muscle activation pattern of the internal oblique, external oblique and rectus abdominis muscles during the “hip knee”, “hip shoulder” and “hip reach” tasks followed the same trend as that found by Butler *et al.* (2008). The internal oblique muscles were recruited to a higher amplitude than the external oblique and rectus abdominis muscles. Additionally, external oblique amplitude of activation was slightly higher than that of the rectus abdominis sites when performing the above tasks (Figure 14 to Figure 16; Appendix C:163). All of the above findings are the same as those reported by Butler *et al.* (2008). The second recruitment strategy observed by Butler *et al.* (2008) involved co-activation between the back extensor sites to a similar amplitude when light external loads were manipulated close to the body. In the present investigation, the lumbar and thoracic erector spinae muscles were activated to similar amplitudes during all loading conditions in the “hip knee” (Figure 12:62 and Figure 13:63), “hip shoulder” and “hip reach” tasks (Appendix C p.163). Similar activation amplitude of muscles has been referred to as “bracing” and has been shown in biomechanical literature to reduce lumbar displacement and increase spinal stability (Vera-Garcia *et al.,* 2006, 2007). Although the present investigation and that performed by Butler *et al.* (2008) elicited similar results, the loads investigated and the type of tasks were, however, different in both investigations. Butler *et al.* (2008) investigated muscle activity associated with a 3kg load and investigation was confined to maximal arm reach. The tasks investigated in this study were, however, similar to those investigated by Butler *et al.* (2008) as the “hip reach” task required increased horizontal reach and the “hip shoulder” task required movement of the arms to an increased distance from the body during task completion. Therefore, similar patterns of muscle activation were not unexpected.

INFLUENCE OF INCREASED LOAD ON GRF AND COP PATTERN

Increased load magnitude had a significant effect on sagittal ground reaction force pattern during both movement phases of the “hip knee” and “hip shoulder” task (Table IV:72). Therefore sagittal ground reaction force pattern was significantly different ($p<0.05$) when the effect of load was negated. Significantly different pattern of sagittal ground reaction force with increased load magnitude could be attributed to the nature of the tasks as both tasks were symmetrical in nature and required movement in the sagittal plane for task completion. A significant interactional effect ($p<0.05$) between condition and lateral ground reaction force pattern was found in the first movement phase of the "hip twist" task (Table IV:72). The absence of significant lateral ground reaction force pattern in the "hip knee", "hip shoulder" and 'hip reach" tasks was expected as tasks were symmetrical in nature requiring movement in the sagittal plane. A significant interactional effect between load and data point was found in vertical ground reaction force in both movement phases of all tasks investigated. Therefore increased load magnitude resulted in significantly different vertical ground reaction force patterns, irrespective of the task performed.

Although sagittal COP movement attributed to increased load magnitude was not significant in all tasks, a significant interactional effect ($p<0.05$) between load and the pattern of COP_x movement occurred in both movement phases of the "hip knee", "hip shoulder" and "hip twist" tasks (Table IV: 72). Therefore, increased load magnitude resulted in greater variability in sagittal movement pattern in the above tasks. In addition, a significant interactional effect between load and COP movement attributed to load in the lateral plane (COP_y) occurred in the first movement phase of the "hip twist" task. Therefore, pattern of right and left movement of the COP was significantly altered in this movement phase.

MOVEMENT COMPONENT OF POSTURAL CONTROL

In order to isolate the muscle activity and ground reaction force associated with body movement, the difference between the static and dynamic condition was calculated at each data point, thereby maintaining muscle activation and ground reaction force pattern. Through this analysis, the effect of lever length, body mass and the effect of the external load on responses was removed, thereby allowing analysis of the muscle activity and ground reaction force associated with body movement. This analysis was performed under each loading condition such that the effect of increased load magnitude on the muscle activity and ground reaction forces associated with body movement could be analysed. It was hypothesised that increased trunk motion would have the same effect on responses irrespective of the load manipulated. Trend in muscle activity attributed to increased body movement differed according to task demands. However, trend in ground reaction forces resulted in similar responses during all tasks and are discussed below.

COMMON TRENDS IN GROUND REACTION FORCES

Table VIII: Trend in ground reaction forces attributed to increased body movement in all tasks investigated.

TASK		F_x	F_y	F_z	CoP_x	CoP_y
"Hip knee"	1					
	2					
"Hip shoulder"	1					
	2					
"Hip twist"	1					
	2					
"Hip reach"	1					
	2					

Where: F_x = Sagittal ground reaction force
F_y = Lateral ground reaction force
F_z= Vertical ground reaction force
COP_x = Movement of the center of pressure in the sagittal plane
COP_y = Movement of the center of pressure in the lateral plane

1 = first movement phase
2 = second movement phase

Movement of the centre of pressure in the sagittal plane (COP_x) was significantly affected ($p<0.05$) by increased load magnitude in both movement phases of all tasks investigated (Table VI:74). Therefore, increased trunk movement under increased external loads resulted in significantly altered body sway and postural stability irrespective of the task performed.

Lateral movement of the COP attributed to increased trunk movement was not significantly different under different loading conditions in both movement phases of the "hip knee" and "hip shoulder task. This was, however expected as both tasks were symmetrical in nature and required sagittal plane movement for task completion.

“MOVEMENT COMPONENT” OF ALL TASKS

“Hip knee” task

Muscle activity

During the first movement phase of the “hip knee” task (lowering the load) the muscle activation attributed to increased body movement of the left external oblique and right internal oblique were significantly different under different load magnitudes (Table V: 73). The absence of a significant effect of load on the remaining muscles during the first movement phase of the “hip knee” task may be due to the nature of movement (lowering the load to knee height). Firstly, this movement occurred in the same direction of gravity thereby reducing the force needed to be overcome in order to control the load. Secondly, during the lowering motion, the external load may have been supported by muscles not investigated in this study, such as the trapezius and biceps muscles. The absence of a significant effect of load in the first movement phase, as well as lower average muscle activity of all muscles during the knee-hip movement phase (Figure 12:62 to Figure 17:66) is in accordance with the literature proposed by Davis *et al.* (1998) which states that lowering is associated with lower muscle activity levels.

All muscles investigated elicited a similar trend in muscle activation during both movement phases of the “hip knee” task as illustrated in Figure 27. Static – dynamic data resulted in greater muscle activation in the 0.8kg loading condition compared to the 4kg loading condition.

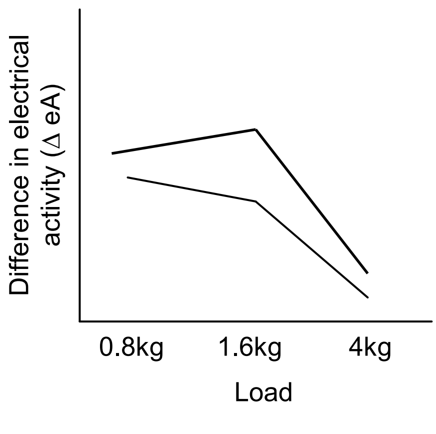


Figure 27: Trends of muscle activity attributed to trunk movement during the “hip knee” task.

In the first movement phase of the “hip knee” task, increased load magnitude significantly affected ($p<0.05$) the muscle activity required for body movement in the left external oblique and right internal oblique muscle (Table V:73). The left external oblique showed an increase from the 0.8kg to 1.6kg condition and then decreased in the 4kg condition. However, the right internal oblique muscle rendered a decrease in muscle activity from the 0.8kg to 4kg loading condition (Figure 27).

The second movement phase of the “hip knee” task resulted in a significant effect of increased load on muscle activity attributed to body movement in all muscles investigated (Table V:73). Therefore, muscle activity associated with body movement was significantly different ($p<0.05$) with changes in load magnitude. The significant effect of load during the second movement phase can be attributed to moving the load against gravity, as this motion would require greater muscle force to accelerate and decelerate the external load. This effect is more pronounced with increased load magnitude, thereby resulting in greater force generation during lifting in comparison with lowering. In the second movement phase, altered muscle activity decreased from the 0.8kg to the 1.6kg and again to the 4kg condition in all muscles, with the exception of the left external oblique muscle (Figure 27). Therefore, when moving the load from hip height to knee height, muscle activation during the static condition of the “hip knee” task was lower than that experienced during both movement phases. This is illustrated in Figure 12(p. 62) to Figure 17 (p. 66).

Muscle activity of the right and left sides differed in response during each movement phase. In the first movement phase, muscle activity attributed to body movement in the left external oblique was significantly affected by increased load magnitude where the right external oblique was not (Table V: 73). The response of these muscles was, however, not significantly different to each other. As the “hip knee” task was symmetrical in nature, this finding can be attributed to possible strength differences between the right and left external oblique. Differences in right and left activation also occurred in the internal oblique muscle during the first movement phase. Additionally, muscle activity associated with body movement of these muscles was not significantly different to each other.

Ground reaction forces

Sagittal COP movement attributed to increased body movement followed a similar trend in the both movement phases of the "hip knee" task as sagittal COP movement attributed to increased trunk movement increased under increased loading conditions (Table VIII:99). In the first movement phase of the "hip knee" task, increased trunk movement resulted in greater body sway with increased external load. In the second movement phase, sagittal COP movement increased in a linear manner from the 0.8kg to 1.6kg loading condition (Table VIII). Therefore, sagittal ground reaction force attributed to increased trunk movement did not increase under light loading conditions. The heavier loading condition (4kg) did, however, result in greater sagittal COP movement with increased trunk movement.

"Hip shoulder" task

Muscle activity

The first movement phase of the "hip shoulder" task was characterised by an absence of a significant effect of load in the right and left external oblique, right rectus abdominis and right latissimus dorsi muscles. Therefore, muscle activity attributed to trunk movement was not significantly different under different external loads in the above muscles. In both movement phases of the "hip shoulder" task, static-dynamic muscle activity was greatest in the 4kg loading condition. Trend of muscle activation of the "hip shoulder" task is illustrated in Figure 28.

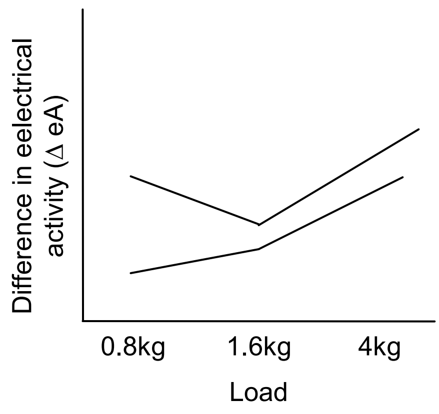


Figure 28: Typical trends of muscle activity attributed to trunk movement during the "hip shoulder" task.

Muscle activity of all significantly affected muscles in the first movement phase, with the exception of the left rectus abdominis, increased from the 0.8kg condition to the 1.6kg condition and further increased to the 4kg condition (Figure 28). This finding combined with the data displayed in Appendix C (Figure 39:166 to Figure 50:170) shows that muscle activation of the static “hip shoulder” task was generally greater than the muscle activity associated with the movement phases of the “hip shoulder” task at each data point.

The effect of load on muscle activity attributed to trunk movement, elicited a greater number of significant differences in the second movement phase of the “hip shoulder” task in comparison to the first movement phase (Table V:73). From this finding it can be deduced that the second movement phase of the “hip shoulder” task (lowering the load to hip height) placed a greater amount of strain on the body in comparison with the first movement phase. This could be attributed to a greater need to control the acceleration and deceleration of the load in the lowering phase. In the second movement phase, muscle activity attributed to trunk movement increased from the 0.8kg to the 1.6kg condition in the right and left lumbar erector spinae, right and left thoracic erector spinae right rectus abdominis, right and left internal oblique and right latissimus dorsi muscle. Decreased muscle activity from the 0.8kg to 1.6kg condition occurred in the left external oblique and left rectus abdominis.

Difference in muscle activity between static and dynamic tasks in the left latissimus dorsi muscle elicited more pronounced changes in muscle activity with increased external load. Additionally, altered data of the left latissimus dorsi muscle was significantly greater ($p<0.05$) than the right latissimus dorsi, further indicating possible strength differences between right and left sides.

Ground reaction forces

Sagittal, lateral and vertical ground reaction force attributed to increased trunk movement was significantly different ($p<0.05$) under different loading conditions in the first movement phase of the “hip shoulder” task. Ground reaction force attributed to increased trunk movement followed the same trend and increased under increased loading conditions (Table VIII:99). From this data it can be proposed that when loads were manipulated from hip height to shoulder height, ground reaction force attributed to trunk movement increased in the sagittal, lateral and vertical planes with increased load magnitude. In the second movement phase, sagittal, vertical and lateral ground reaction force attributed to increased trunk movement were not significantly different under different loading conditions (Table VI:74). This indicates that the first movement phase of the hip shoulder task was more substantially influenced by increased trunk movement under different external loads in comparison with the first movement phase.

Sagittal COP movement attributed to increased trunk movement followed the same trend in both movement phases of the "hip shoulder" task and increased under increased loading conditions (Table VIII:99). From this data it can be concluded that COP movement attributed to increased trunk movement increased with increased load magnitude. Further, it can be stated that body sway was significantly affected by increased load magnitude during both movement phases of the "hip shoulder' task.

"Hip twist" task

Muscle activity

In contrast to the other tasks investigated, muscle activation attributed to trunk movement did not remain constant in both movement phases of the "hip twist" task. Trends of muscle activation during the first movement phase are illustrated below.

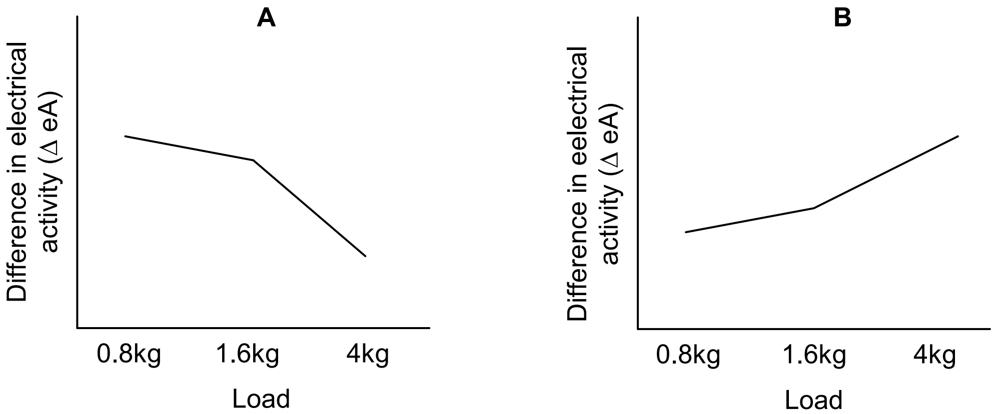


Figure 29: Trends in muscle activity attributed to trunk movement during the first movement phase of the "hip twist" task.

In the first movement phase of the “hip twist” task, muscle activity attributed to trunk movement in the left lumbar erector spinae, right thoracic erector spinae and right internal oblique decreased from the 0.8kg to the 4kg condition as shown in Figure 29A. Therefore, average static muscle activation of the “hip twist” task was of a lower magnitude than the muscle activity associated with the first movement phase in these muscles. This created a downward trend in static-dynamic muscle activity with increased load magnitude. Activity of the left external oblique, left rectus abdominis and left internal oblique followed the opposite trend, shown in Figure 29(B). These muscles displayed an increase in muscle activity attributed to trunk movement with increased load magnitude. From the trend in muscle activity of the left external oblique, left rectus abdominis and left internal oblique, it can be concluded that static muscle activity of the “hip twist” task was greater than the muscle activity associated with the first movement phase under each loading condition. During the first movement phase, subjects twisted towards the right hand side. The increased activity of the left muscles under increased load indicates possible strength differences between the right and left sides.

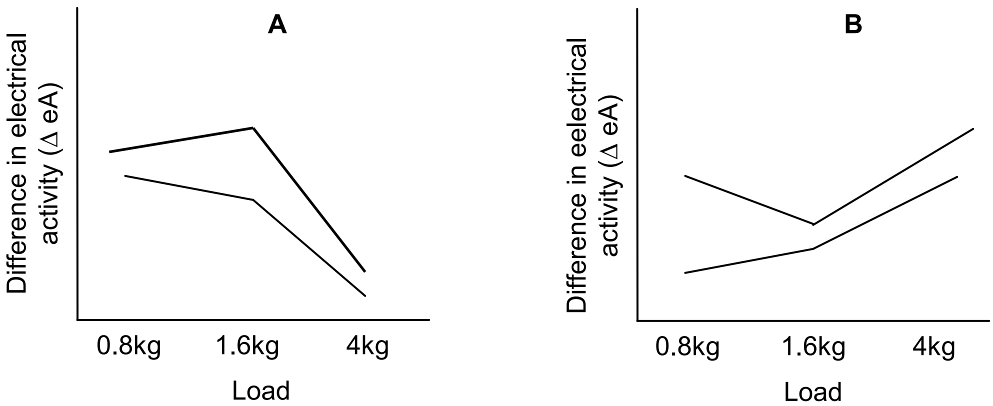


Figure 30: Trends in muscle activity attributed to trunk movement during the second movement phase of the “hip twist” task.

In the second phase of movement of the “hip twist” task, the right and left lumbar erector spinae and right and left thoracic erector spinae followed a similar trend in muscle activation and is illustrated in

Figure 30(A). All of the above muscles with the exception of the right thoracic erector spinae showed an increase in muscle activity attributed to trunk movement from the 0.8kg to the 1.6kg condition. Therefore, trunk movement resulted in additional strain on the musculoskeletal system under 0.8kg to 1.6kg loads in these muscles during the second movement phase of the “hip twist” task. Muscle activation of the right thoracic erector spinae decreased between the two loading conditions. Muscle activity attributed to body movement was lower in the 4kg condition compared to the 0.8kg condition in all of the above muscles. Thus, increased body movement resulted in lower muscle activity at increased loads when tasks were performed dynamically.

The left external oblique, left rectus abdominis, left internal oblique and left latissimus dorsi muscles followed the trend displayed in Figure 30(B) and muscle activity attributed to body movement was greater in the 4kg condition than in the 0.8kg condition. Muscle activity attributed to increased trunk movement increased with increased load magnitude in the left external oblique, left rectus abdominis, left internal oblique and left latissimus dorsi muscles. Thus, muscle activity associated with the second movement phase of the “hip twist” task was lower than that of the static “hip twist” task in these muscles. Therefore, performing the dynamic component of the “hip twist” task towards the non-dominant side resulted in additional strain on the musculoskeletal system with increased load magnitude.

Different trends in muscle activation associated with body movement in the “hip twist” task were expected due to the nature of the task and the function of the muscles investigated. Greater muscle activity in the left external oblique, rectus abdominis, internal oblique and latissimus dorsi with increased load magnitude can be attributed to the nature of the task as the trunk rotated to the left in the second movement phase of the “hip twist” task. These muscles were also all located on the left side of the body and increased activation of these muscles is therefore expected.

Ground reaction forces

Ground reaction force attributed to increased trunk movement in all planes was significantly different under different load magnitudes in both movement phases of the “hip twist” task with the exception of sagittal ground reaction force in the first movement phase (Table VI:74). Additionally, sagittal and lateral COP movement attributed to increased trunk movement was significantly different ($p<0.05$) under different external loads in both movement phases of the “hip twist” task. The trend of lateral ground reaction force differed between the first and second movement phase but vertical ground reaction force and sagittal and lateral COP movement followed the same trend during the first and second movement phase (Table VIII:99).

In the first movement phase, lateral ground reaction force attributed to increased body movement decreased from the 0.8kg to 1.6kg loading condition and subsequently increased to greater levels than the 0.8kg condition in the 4kg condition (Table VIII:99). Hence, twisting towards the dominant side resulted in decreased lateral movement under light loading conditions but greater lateral ground reaction force under higher external loads. The opposite trend occurred in the second movement phase as lateral ground reaction force attributed to body movement increased under light loading conditions and decreased under heavier loading conditions (Table VIII:99). In both movement phases of the "hip twist" task, sagittal movement of the COP associated with increased trunk movement was lower at higher loads in comparison with lighter loads (Table VIII:99). This indicates a reduction in anterior/ posterior COP movement with increased load magnitude, thereby creating a more stable movement. Lateral COP movement attributed to increased trunk motion followed the same trend in both movement phases and increased with increased load magnitude (Table VIII:99). The presence of significance in the lateral plane and absence of significance in the sagittal plane was expected due to the nature of the task. From this data it can be proposed that lateral ground reaction force attributed to increased trunk movement increased with increased load magnitude when twisting towards the dominant side and decreased under heavier loading conditions when twisting towards the non-dominant side. Both movements were accompanied by increased lateral COP movement and decreased sagittal COP movement with increased load magnitude.

Therefore, increased asymmetrical trunk movement resulted in significantly altered ground reaction forces under different loading conditions. These findings are also in support of literature which proposes a decrease in trunk strength (Marras and Mirka, 1989; Ferguson *et al.,* 1992) and more complex trunk motion patterns in asymmetrical lifts (Ferguson and Marras, 1992; Ferguson *et al.,* 1992; Allread *et al.,* 1996).

“Hip reach” task

Muscle activity

During both movement phases of the “hip reach” task, muscle activity attributed to body movement was significantly different (p<0.05) under different loading conditions in all muscles investigated (Table V:73). Additionally, the muscle activity attributed to body movement was greatest in the 4kg condition (Figure 31).

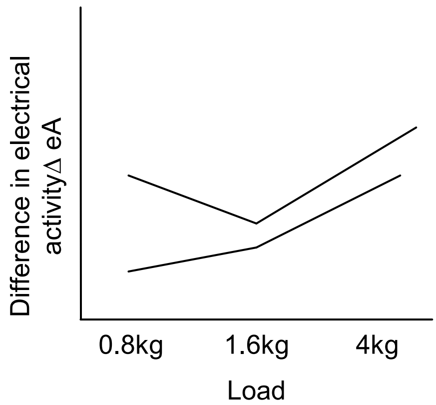


Figure 31: Typical trend of muscle activity attributed to trunk movement in the “hip reach” task.

In the first movement phase, nine of the 12 muscles investigated increased from the 0.8kg to the 1.6kg condition. These included the right lumbar erector spinae, right and left thoracic erector spinae, right and left rectus abdominis and right and left latissimus dorsi muscles. This finding indicates that the muscle activity associated with the static condition of the “hip reach” task elicited greater muscle activity than the dynamic conditions under all investigated loading conditions in the above muscles (Appendix C Figure 63:177 to Figure 74:181). Muscle activity decreased from the 0.8kg to 1.6kg loading condition in the left lumbar erector spinae, right and left external oblique and the left internal oblique muscles. Muscle activity attributed to trunk movement increased in the 4kg condition to greater levels than that of the 0.8kg and 1.6kg condition in all muscles. From this data it can be concluded that muscle activity associated with the static “hip reach” task under the 4kg condition was greater than that of both movement phases in the “hip reach” task. As such, performing tasks dynamically did not decrease the muscle activity associated with body movement when performing the 4kg condition.

Muscle activity attributed to trunk movement of the right lumbar erector spinae was significantly greater ($p<0.05$) than the left lumbar erector spinae during both movement phases of the “hip reach” task. This can be attributed to the difference in response of the right and left muscle from the 0.8kg to 1.6kg loading condition (Table V: 73). The decrease in muscle activity of the left lumbar erector spinae from the 0.8kg to 1.6kg condition in both movement phases could be due to an increased reliability on the right muscle, thereby resulting in significantly different muscle activation. In the first movement phase, muscle activity of the left internal oblique was significantly greater ($p<0.05$) than the right internal oblique. As this task was symmetrical, this difference in muscle activation may be attributed to strength differences.

In contrast to the other tasks investigated, the presence and trend of muscle activity between loading conditions was the same in both movement phases of the “hip reach” task. These findings indicate that when manipulating loads with extended horizontal reach, the direction of load manipulation does not impact the trend of trunk muscle activity attributed to body movement.

Ground reaction forces

During the first movement phase of the “hip reach” task, vertical ground reaction force and sagittal COP movement attributed to increased trunk motion was significantly different under different external loads (Table VI:74). The trend of these variables was similar as ground reaction forces under the 4kg condition were greater than that of the 0.8kg condition (Table VIII:99). Sagittal COP movement did, however, decrease from the 0.8kg to 1.6kg condition revealing smaller sagittal deviations when manipulating light external loads in conditions of extended horizontal reach. The absence of a significant effect of increased trunk movement on sagittal plane ground reaction force was not expected as increased trunk movement in this task resulted in greater sagittal plane movement.

In the second movement phase of the “hip reach” task, sagittal, lateral and vertical ground reaction force, as well as sagittal and lateral COP movement attributed to increased trunk movement was significantly different under different loading conditions (Table VI:74). In addition, all ground reaction force variables decreased from the 0.8kg to 1.6kg condition resulting in lower sagittal, lateral movement with increased external load when manipulating loads from an extended horizontal reach to the hips (Table VIII:99). Lateral ground reaction force and sagittal movement of the COP did, however, increase from the 1.6kg to 4kg condition but did not exceed that of the 0.8kg condition (Table VIII:99). The presence of significance in lateral ground reaction force during the second movement phase was not expected as the "hip reach' task was symmetrical in nature. However, according to this data, subjects adopted a greater lateral movement pattern in the second movement phase of the "hip reach" task. Significantly greater vertical ground reaction force attributed to increased trunk movement in the second movement phase can be attributed to subjects pushing off of the box to return to an upright standing position. This movement would alter vertical ground reaction force and this effect would be different under different loading conditions. In the second movement phase of the "hip reach" task, lateral COP attributed to increased trunk movement decreased under increased loading conditions. Thus, increased load magnitude resulted in smaller lateral COP deviations attributed to increased body movement. From this data, the second movement phase of the "hip reach" task resulted in greater differences in weight shift patterns with increased load magnitude in comparison with the first movement phase of the "hip reach" task.

INFLUENCE OF INCREASED BODY MOVEMENT ON MUSCLE ACTIVITY PATTERN

Trunk recruitment patterns have shown to differ when loads are increased (Fathallah *et al.,* 1998; de Looze *et al.,* 1999; Davis and Marras, 2000). This research is in support of this literature as a significant interactional effect ($p<0.05$) between load and the pattern of muscle activity attributed to body movement was found in all muscles and all tasks investigated, with the exception of the right latissimus dorsi in the "hip twist" task, the right rectus abdominis in the "hip shoulder" task and the left rectus abdominis in the "hip knee" task (Table V:73). Therefore, trunk recruitment patterns differ with increased external load magnitude irrespective of the task performed. The presence of significant differences in muscle activity pattern in the "hip twist" task is in accordance with the findings of Ferguson and Marras (1992), Ferguson *et al.* (1992) and Allread *et al.* (1996) which state that asymmetrical lifting is associated with the development of more complex trunk motion patterns.

Muscle activation pattern associated with increased body movement was significantly different between right and left muscles in some tasks. In the “hip shoulder” task, pattern of muscle activation of the right and left latissimus dorsi were significantly different to each other ($p<0.05$) during both movement phases. This is indicative of greater variability in muscle recruitment of the right and left latissimus dorsi muscles. This can be attributed to an increased contribution of the latissimus dorsi muscles during both movement phases which would increase with increased external load magnitude. Muscle activity pattern of the right and left rectus abdominis were also significantly different to each other in the second movement phase of the “hip shoulder” task. In the “hip reach” task, muscle activation of the rectus abdominis and internal oblique in the first movement phase and lumbar erector spinae in the second movement phase differed significantly between right and left muscles. From the above findings it can be proposed that increased load magnitude results in a significantly different muscle recruitment pattern between right and left sides in symmetrical tasks which require extended horizontal reach.

INFLUENCE OF INCREASED BODY MOVEMENT ON GROUND REACTION FORCE PATTERN

Pattern of sagittal ground reaction force attributed to increased body movement was significantly different ($p<0.05$) under different external loads in all tasks, with the exception of the first movement phase of the ‘hip reach” task (Table VI:74). Therefore, increased external load resulted in significantly different anterior posterior weight shift patterns, irrespective of the task performed. A significant interactional effect ($p<0.05$) between load and lateral ground reaction force pattern occurred in the first movement phase of the “hip knee” and “hip shoulder” tasks, both movement phases of the “hip twist” task and the second movement phase of the “hip reach” task. Therefore, lateral ground reaction force pattern attributed to increased trunk movement was significantly different under different external loads during the above mentioned movement phases. A significant interactional effect ($p<0.05$) occurred in vertical ground reaction forces in both movement phases of all tasks investigated, indicating different patterns of vertical loading with increased body movement under different external loads (Table VI:74).

Pattern of sagittal COP movement attributed to increased trunk movement was also significantly altered with increased load magnitude in all tasks investigated (Table VI:74). Thus, increased trunk movement resulted in significantly different body sway patterns under different external loads, which may be indicative of greater postural instability. A significant interactional effect ($p<0.05$) between load and lateral COP movement attributed to increased trunk movement occurred in the first movement phase of the “hip knee” and “hip shoulder” task, both movement phases of the “hip twist” task and the second movement phase of the “hip reach” task. This finding was not expected in the symmetrical tasks but could be indicative of a slight loss of balance in a lateral direction during these tasks. This finding may also suggest an increased shift of weight onto one leg during task completion.

BODY DISCOMFORT

Nielsen *et al.* (1998) stated that there is a trade off between the back and shoulder muscles when lifting from different heights. The body discomfort ratings of all tasks support this literature as greater discomfort in the posterior trunk was accompanied by lower discomfort ratings in the upper limbs and neck and visa versa. This trade off occurred under both static and dynamic conditions as well as under all loading conditions and is illustrated in Figure 23 (p.76). Average static body discomfort (Figure 24:77) associated with the 0.8kg loading condition was not significantly different between tasks under the same loading condition. Therefore, according to this data, the task performed does not significantly influence the degree of discomfort experienced when manipulating external loads of 0.8kg.

“Hip knee” task

Average static body discomfort associated with the 1.6kg and 4kg conditions of the “hip knee” task was significantly lower ($p<0.05$) than the discomfort experienced in the “hip reach” task under both loading conditions (Figure 24:77). This can be attributed to the posture adopted during the static conditions of the above tasks. During the static condition of the “hip knee” task, the mass of the body and load may have been supported by the spinal ligaments and intervertebral discs, thereby reducing muscle activity and subsequent body discomfort. This is in contrast to the static condition of the “hip reach” task in which subjects were required to adopt a posture which increased the mechanical disadvantage of the body (Jorgensen *et al.,* 2003) which subsequently resulted in greater stain.

Discomfort experienced during the dynamic condition of the “hip knee” task was greater than that of the static condition under all loading conditions (Figure 24:77 and Figure 25:78). This was particularly evident in the posterior and anterior trunk as illustrated in Figure 23 (p.76). Increased discomfort in these regions can be attributed to the function of the trunk muscles which flex and extend the trunk in dynamic conditions (Tortora and Grawbowski, 2003). From this data it can be concluded that the dynamic condition of the “hip knee” task was associated with greater strain in the anterior and posterior trunk, indicating greater muscle activity during the dynamic condition. Body discomfort of the dynamic component of the “hip knee” task was significantly greater ($p<0.05$) that the discomfort experienced in the “hip twist” tasks under all loading conditions (Figure 25:78). Therefore, the dynamic condition of the “hip knee” task was perceived to be more strenuous than the “hip twist” task, irrespective of the external load magnitude. Increased level of trunk motion has been associated with increased agonist and antagonist muscle activity (Marras *et al.,* 1984; Cresswell and Thorstensson, 1989, Granata and Marras 1995b; Gallagher, 1997). The perceptual responses of the “hip knee” task are in accordance with this literature as a greater trunk motion was associated with increased discomfort in the anterior and posterior trunk.

“Hip shoulder” task

Average static body discomfort of the “hip shoulder” task in the 4kg loading condition (Figure 24:77) was significantly greater ($p<0.05$) than the discomfort of both the “hip knee” and “hip twist” tasks performed with the same load. This can be attributed to the posture adopted during the static condition of the “hip shoulder” task. Subjects were required to hold the external load at shoulder height with the arms extended, thereby increasing mechanical disadvantage, resulting in greater muscle activity to counteract the load. This is supported by the regions of discomfort during this condition as discomfort in the upper limbs and neck was markedly greater in the static “hip shoulder” task in comparison with the “hip knee” and “hip twist” tasks (Figure 23:76). Although both the “hip reach” and “hip shoulder” tasks resulted in increased mechanical disadvantage, the discomfort experienced in the static 4kg condition of the “hip shoulder” task was significantly lower ($p<0.05$) than that of the “hip reach” task under the same loading condition. This suggests that the static condition of the “hip shoulder” task was perceived as less strenuous when compared to the static component of the “hip reach” task.

Body discomfort during the static and dynamic conditions of the “hip shoulder” task was similar under all loading conditions. Additionally, regions of discomfort did not change markedly between the static and dynamic conditions (Figure 23:76). Average body discomfort of the dynamic component of the “hip shoulder” task resulted in lower discomfort levels than the “hip knee” task under all loading conditions. This difference was significant ($p<0.05$) in the 0.8kg loading condition. Total dynamic discomfort in the upper limbs and neck was greater in the “hip shoulder” task, where discomfort in the posterior trunk was greater in the “hip knee” task. Total average discomfort may have been lower in the “hip shoulder” task due to the absence of rating in the lower limbs and anterior trunk. This would affect total discomfort during the “hip shoulder” task and result in lower discomfort ratings. Dynamic discomfort associated with the 4kg condition was significantly greater in the “hip shoulder” task in comparison with the “hip twist” task (Figure 25:78). In addition, total discomfort was greater in the upper limbs and neck in the “hip shoulder” task than in the “hip twist” task (Figure 23:76). Greater total discomfort in the upper limbs and neck during the “hip shoulder” task was expected as the arms were the primary movers in this task. Average rating was, however, similar in this region under all loading conditions in the “hip shoulder” and “hip twist” tasks (Figure 23:76).

“Hip Twist” task

The static condition of the “hip twist” task was perceived to be the least demanding as discomfort during this condition was significantly lower ($p<0.05$) than the “hip shoulder” task under the 4kg loading condition (Figure 24:77). Additionally, discomfort experienced during the static “hip twist” task performed under the 1.6kg and 4kg load was significantly lower ($p<0.05$) than that of the “hip reach” task under the same loading conditions. Further, the “hip twist” task was associated with the smallest discomfort ratings under all loading conditions. The lower discomfort in the "hip twist" task can be attributed to the characteristics of the task as subjects were required to twist the body in the lateral plane at a constant working height. The "hip twist" task also did not result in increased lever length during task completion thereby reducing internal loading and associated discomfort

Discomfort associated with the dynamic condition of the "hip twist" task was slightly greater than the discomfort experienced during the static condition (Figure 24: 77 and Figure 25:78). Total and average discomfort rating in the anterior and posterior trunk during the "hip twist" task was greater in the dynamic condition under all loading conditions. Discomfort did, however, decrease from the static to dynamic condition in the upper limbs and neck (Figure 23:76). Consequently, performing twisting tasks such as those investigated resulted in lower discomfort in the upper extremities accompanied by increased discomfort in the posterior and anterior trunk. Average discomfort of all body regions of the dynamic "hip twist" task was significantly lower ($p<0.05$) in the 0.8kg and 1.6kg condition in comparison with the "hip knee" task (Figure 25:78). Additionally, discomfort of the 1.6kg condition was significantly lower than that of the same loading condition of the "hip reach" task. Further, total average discomfort was significantly lower ($p<0.05$) during the 4kg condition of the "hip twist" task in comparison with the "hip knee", "hip shoulder" and "hip reach" tasks. Therefore, under all loading conditions, the dynamic "hip twist" task was perceived as the least strenuous.

"Hip reach" task

The static condition of the "hip reach" task resulted in the greatest average body discomfort under all loading conditions (Figure 24:77). This was expected as the static condition of the "hip reach" task required static muscle activation in conditions of increased lever length. Body discomfort experienced in the static condition of the "hip reach" task performed under to 4kg loading condition was significantly greater ($p<0.05$) than the discomfort experienced under the same loading conditions in all tasks investigated (Figure 24:77). Thus, the static component of the "hip reach" task performed under the 4kg loading condition was perceived to be the most strenuous task, resulting in greatest discomfort.

The body discomfort associated with the dynamic component of the “hip reach” task in the 4kg condition was significantly greater ($p<0.056$) than discomfort during the “hip twist” task under the 4kg and 1.6kg loading condition (Figure 25:78). Therefore, with increased external load, the dynamic component of the “hip reach” task was perceived as the most demanding task. This could be attributed to the nature of the “hip reach” task as discussed previously.

INTEGRATED DISCUSSION

LOAD COMPONENT OF POSTURAL CONTROL

The trend of muscle activation attributed to load remained similar across all muscles and tasks investigated in response to increased load magnitude. Muscle activity attributed to load did not increase in a linear manner with the exception of the left rectus abdominis in the second movement phase of the “hip twist” task and the right thoracic erector spinae in the first movement phase of the “hip reach” task. Consequently, in the above muscles, increased load magnitude did not place additional strain on the musculoskeletal system. The linear function associated with these muscles did, however, only occur between the A and 4B conditions. Muscle activity attributed to load increased in the 3C condition to levels greater than that of the A condition for all muscles investigated. Consequently, higher load magnitudes result in additional strain on the body in excess of that imposed by task demands. Decreased muscle activity from the A to 4B condition in specific muscles and tasks (Table III:71) suggests a degree of movement efficiency when manipulating light external loads.

In the second movement phase of the “hip knee” task, at higher load magnitudes, increased muscle activity attributed to load was accompanied by a decrease in sagittal ground reaction force. As such, increased muscle activity may be a result of decreasing sagittal ground reaction force movement (Table VII:81). The same principle can be applied to the second movement phase of the “hip twist” task and the second movement phase of the “hip reach” task as sagittal ground reaction force attributed to increased load decreased with increased load magnitude. Increased muscle activity attributed to load was also accompanied by a decrease in lateral ground reaction force and decreased sagittal COP movement in the first and second movement phase of the "hip twist" task respectively. Additionally, increased load magnitude resulted in greater total body discomfort ratings in all tasks investigated (Figure 23:76). This finding is in accordance with the increased muscle activity attributed to load under heavier loading conditions.

It has been established that antagonist co-activation is necessary to maintain the mechanical stability of the spine (Cholewicki and McGill, 1996; Granata and Wilson, 2001; Vera-Garcia *et al.,* 2007) which is particularly important in tasks which require little physical effort (Cholewicki and McGill, 1996; Cholewicki *et al.,* 1997). Antagonist recruitment strategies which occurred in this investigation were in accordance with those observed by Butler *et al.* (2008), particularly in the "hip shoulder" and "hip reach" tasks.

The findings of this investigation support those of Houdijk *et al.* (2009) who found that balance perturbations resulted in increased energy consumption. The same authors attributed this to either extra muscular effort required to alter the execution of the primary motor task or from increased muscular effort, which is relevant to this investigation. Houdijk *et al.* (2009) further propose that when maintaining a static posture or moving in a cyclic pattern (such as that which occurs during repetitive lifting) increasing stability may require additional muscular effort to reduce movement variability. The current research supports this literature and the concept of increased physical demand attributable to postural control as increased load magnitude resulted in additional muscle activity in excess of that imposed by task demands. Due to the non-linear increase in muscle activity associated with load when the effect of load was negated, it can be concluded that performing tasks with increased load magnitude placed additional strain on the body, in excess of that imposed by task demands. This increase in muscle activity may be necessary to maintain spinal stability and postural equilibrium. Therefore, lifting guidelines may underestimate muscle activity and spinal load loading in tasks similar to those investigated due to the additional strain on the body as a result of increased load magnitude.

The pattern of muscle activity associated with increased load magnitude was significantly different ($p<0.05$) under different load magnitudes even when the effect of load was negated during the “hip knee”, “hip shoulder” and “hip twist” tasks. This was accompanied by a significant interactional ($p<0.05$) effect between load and altered pattern of COP movement in the sagittal plane. Therefore, even when the effect of load was negated, higher loading conditions resulted in altered muscle activity pattern, as well as altered pattern of body sway. Although the movement of the COP in the sagittal plane was not significantly different ($p<0.05$) when the effect of load was negated, a significant interactional effect ($p<0.05$) between condition and muscle activation pattern was present in of the majority of investigated muscles during all tasks (Table III:71). According to electromyographic and force plate data, muscle activity and ground reaction force attributed to load were most affected in the first movement phase of the “hip reach” task as all muscles were significantly altered. This was accompanied by significantly altered sagittal, lateral and vertical ground reaction force. Additionally, the “hip reach” task was perceived to be the most strenuous task.

MOVEMENT COMPONENT OF POSTURAL CONTROL

Literature states that strength is reduced by up to 30% when tasks are dynamic in nature and as a result, at the same exertion level, dynamic movements are closer to the muscular tolerance level in comparison with static exertions (Granata and Marras, 1995a, b; Gallagher, 1997; Dempsey *et al.,* 1998). The electromyographic data from both movement phases of the “hip knee” task supports this notion as static muscle activity was lower than the average muscle activity of both movement phases (Figure 12:62 to Figure 17:66). This occurred in all muscles investigated. In contrast, static muscle activation of the “hip shoulder” task was greater than average muscle activity of the movement phases in the right lumbar erector spinae and right thoracic erector spinae, rectus abdominis, internal oblique, latissimus dorsi muscles. In the “hip shoulder” task, the only muscle in compliance with the above literature was the right external oblique muscle in the 0.8kg and 1.6kg conditions (Appendix C Figure 43:167). Therefore, when tasks are performed between hip height and shoulder height, static tasks may place greater strain on the trunk muscles when compared to dynamic movements. This trend may be a result of the lower degree of trunk flexion and extension in the “hip shoulder” task compared to the “hip knee” task. Additionally, the static condition of the “hip knee” task occurred with the box placed at knee height.

At this point of trunk flexion, the mass of the trunk and the external load may be counteracted by the spinal ligaments and intervertebral discs (McGill and Kippers, 1994) thereby reducing muscular effort and muscle activation level in the static condition of the “hip knee” task. Further, the static condition of the “hip shoulder” task placed the body at a greater mechanical disadvantage as the arms were extended at shoulder height. The posture adopted during this task would require greater muscle activity to overcome the external force and additional strain as a result of an increased lever arm (Jorgensen *et al.,* 2003).

The “hip reach” task was also in contrast to the above literature as static muscle activation was greater than average muscle activation of both movement phases in the right lumbar erector spinae, right thoracic erector spinae, right rectus abdominis, right internal oblique and right latissimus dorsi muscles (Appendix C Figure 63:177 to Figure 74:181). It should, however, be noted that peaks in muscle activity pattern in both the “hip reach” and “hip shoulder” task, did on occasion, exceed the average static muscle activity of each task. From these findings it can be proposed that tasks which require increased lever length, and therefore place the body at a greater mechanical disadvantage, contradict the literature. In such tasks, static muscle activity at the most strenuous point in movement may be closer to muscle tolerance levels in comparison with dynamic tasks.

Muscle activity attributed to increased trunk movement was significantly different under different external loads, indicated by the presence of significance in Table V (p.73). In the “hip knee” task, muscle activity associated with increased body movement decreased with increased external load. This was accompanied by an increase in sagittal COP movement with increased load (Table VIII:99). Therefore, in the “hip knee” task, although muscle activity associated with increased body movement decreased, body sway attributed to increased trunk movement, increased with increased load magnitude. This is indicative of decreased postural stability with increased load magnitude. Consequently, dynamic movement between hip height and knee height resulted in decreased muscle activity and decreased postural stability with increased load magnitude. Performing dynamic tasks with increased load was also perceived to be more demanding compared to static tasks (Figure 23:76).

Increased body movement in the “hip shoulder” task resulted in an increase in muscle activity associated with body movement with increased load magnitude (Figure 28:102). This was accompanied by increased sagittal, lateral and vertical ground reaction force attributed to increased body movement with increased load magnitude (Table VIII: 99). Sagittal COP movement attributed to body movement also increased with increased load magnitude. Therefore, dynamic lifts in the “hip shoulder” task resulted in greater postural instability, greater sagittal, lateral and vertical ground reaction forces as well as increased muscle activity under different loading conditions. Perceived discomfort also increased from the static to dynamic condition in the “hip shoulder” task (Figure 23:76).

The trend of ground reaction force and muscle activity attributed to body movement differed between the two movement phases of the “hip twist” task. Although the “hip twist” task was perceived to be the least demanding (Figure 23:76), muscle activity and ground reaction forces attributed to increased body movement were significantly different ($p<0.05$) under different loading conditions. The second movement phase of the “hip twist” task was associated with greater variation in sagittal ground reaction force in comparison with the first movement phase (Table VI:74) as well as an increased prevalence of significantly altered muscle activity under different external loads (Table V:73). Therefore, performing twisting movements towards the non-dominant side resulted in greater changes in weight shift patterns and muscle activity under different loading conditions. Body sway attributed to increased body movement did, however, decrease with increased load in the second movement phase of the “hip twist” task (Table VIII:99). Consequently, increased muscle activity attributed to increased body movement may have served to decreased sagittal COP movement, resulting in greater stability.

Muscle activity associated with increased body movement was significantly different under different loading conditions in all muscles investigated during both movement phases of the “hip reach” task. Significantly different muscle activity attributed to external load in the first movement phase of the “hip reach” task was accompanied by greater postural instability from the 0.8kg to 4kg condition (Table VIII:99). Ground reaction forces attributed to increased body movement were, however, altered to a greater degree in the second movement phase of the "hip reach" task as ground reaction force attributed to increased body movement were significantly different ($p<0.05$) under different external loads in all ground reaction force variables (Table VIII:99). In the second movement phase of the "hip reach" task, lateral COP attributed to increased body movement decreased under increased loading conditions and therefore less lateral movement occurred during this movement phase. As a result, the increased muscle activity associated with increased load may have served to decrease sagittal ground reaction force and lateral movement of the COP during the "hip reach" task.

CONCLUSION

Statistical analyses performed on manipulated data indicate that performing tasks with increased load magnitude and with increased trunk movement results in additional strain on the musculoskeletal system, over and above that imposed by task demands. A degree of movement efficiency did occur between the loads of 0.8kg and 1.6kg in which muscle activity and ground reaction forces attributed to load decreased with increased external load. This, however, did not occur under heavier loading conditions as the 4kg condition resulted in greater muscle activity attributed to load than the 0.8kg and 1.6kg condition. Muscle activity and ground reaction forces attributed to increased body movement generally placed additional strain on the body in tasks which involved an extended horizontal reach. This research shows that performing tasks with an added external load and performing tasks with a greater degree of trunk motion can result in a greater demand on the musculoskeletal system than previously thought. This may cause guidelines to underestimate risk and may be a contributing factor as to the rise in low back disorders despite the implementation of work guidelines.

CHAPTER VI
SUMMARY, CONCLUSIONS AND RECOMMENDATIONS

INTRODUCTION

The most common musculoskeletal disorders are low back disorders (Marras, 2000) and while most research has tended to ignore the impact of postural muscle control, it has been shown fairly recently that risk of occupationally-related low back disorders may be associated with spinal stability (Granata and Wilson, 2001). Despite the fact that vast amounts of research has been performed with the aim of reducing the risks of manual materials handling and the development of low back disorders, the effectiveness of existing guidelines in reducing these risks is questionable as low back disorders remain highly prevalent (Marras, 2000). One explanation could be the fact that these guidelines do not take into account the postural control mechanisms which are necessary for safe and efficient manual handling (Kollmitzer *et al.,* 2002). It has been proposed that the increased physiological demand associated with postural instability (Houdijk *et al.,* 2009) may be a result of increased muscular effort required to maintain postural equilibrium (Houdijk *et al.,* 2009).

Although it has been shown that postural control is task dependent, only a few studies have sought to investigate the influence of task demands on the mechanisms of postural control (Streepy and Angulo-Kinzler, 2002; Amiridis *et al.,*2003; Prioli *et al.,* 2006). This study aimed to establish the effect of increased load mass and increased trunk motion on muscle activity and ground reaction forces. It was contended that increased external load and increased trunk movement places additional strain on the body, in excess of that imposed by task demands.

SUMMARY OF PROCEDURES

Experimentation took place in a laboratory setting in the Human Kinetics and Ergonomics Department of Rhodes University. Subjects took part in one testing session of approximately an hour and a half. Twenty four male and 24 female student volunteers between the ages of 18 and 25 participated in this investigation. All subjects were free from low back pain and injury and were physically active, participating in exercise three times per week.

Muscle activity of the upper and lower erector spine, internal oblique, external oblique, rectus abdominis and latissimus dorsi muscles were analysed using surface electromyography (EMG) throughout the testing session. EMG data was normalised to maximum voluntary contractions which were performed prior to task completion. Ground reaction forces were also measured throughout testing through the use of a force platform. Postural stress was analysed using the Body Discomfort Scale and Map developed by Corlett and Bishop (1976). Areas of discomfort were identified and rated on a scale of one to ten at the end of each condition.

Subjects were required to perform four load manipulation tasks. Each task comprised of a static and dynamic component and was completed under three loading conditions (0.8kg, 1.6kg and 4kg). In the static condition, subjects were required to maintain a specific static posture for 10 seconds. For the dynamic condition, subjects were required to move a box within designated areas for the duration of one minute. Movement pace was set at one transfer every three seconds such that the box was moved from and back to its original position in six seconds. Consequently 10 lift and replace cycles were performed in each dynamic condition. A metronome was used to control lifting pace during the dynamic condition and a rest period of one minute was provided between each condition in order to avoid fatigue.

SUMMARY OF RESULTS

Differential analysis was used to isolate the proposed “load component” and “movement component” of postural control. The proposed load component of postural control was isolated by manipulating data such that the effect of load was negated. Additionally, the difference between muscle activity and ground reaction forces elicited during the static and dynamic condition was used to isolate the proposed “movement component” of postural control. Body discomfort experienced during all tasks was also recorded and analysed.

Load component of postural control

Muscle activity attributed to load followed a similar trend during all tasks and did not increase in a linear manner. A degree of movement efficiency occurred between the 0.8kg and 1.6kg condition in which muscle activity attributed to load decreased. Muscle activity attributed to load did, however, increase from the 0.8kg to 4kg loading condition in all muscles and for all tasks investigated. Therefore, increased load placed additional strain on the body in addition to that of task demands for all tasks investigated. A significant interactional effect was found between condition and data point in the majority of muscles during both movement phases of all tasks investigated. Therefore, increased load magnitude significantly altered muscle activity pattern during the completion of all tasks.

When the effect of load was negated, vertical ground reaction forces were significantly different ($p<0.05$) in all tasks with the exception of the second movement phase of the "hip shoulder" task. Sagittal ground reaction force was significantly different ($p<0.05$) with increased external load in the second movement phase of the "hip knee" task, the first movement phase of the "hip shoulder" and "hip reach" tasks and both movement phases of the "hip twist" task. The trend of sagittal ground reaction force was also similar in all cases with the exception of the first movement phase of the "hip twist" task and increased from the A to 4B condition and decreased from the 4B to 3C condition. Lateral ground reaction force was significantly altered in both movement phases of the "hip twist" task and the first movement phase of the "hip reach" task. The trend of lateral ground reaction force decreased from the A to 4B condition and increased in the 3C condition in the second movement phase of the "hip twist" task and the first movement phase of the 'hip reach" task. The opposite trend occurred in first movement phase of the "hip twist" task. Sagittal centre of pressure movement was significantly different ($p<0.05$) when the effect of load was negated in the second movement phase of the "hip shoulder" and "hip twist" tasks. The trend of centre of pressure movement differed between the two tasks and increased with increased load in the "hip shoulder" task. Lateral centre of pressure movement was significantly different ($p<0.05$) when the effect of load was negated in both movement phases of the "hip shoulder" task and the second movement phase of the "hip twist" task. Trend of lateral centre of pressure movement was the same during the above tasks and decreased from the A to 4B condition and increased in the 3C condition.

The pattern of vertical ground reaction force was significantly different ($p<0.05$) when the effect of load was negated in both movement phases of all tasks investigated. A significant interactional effect ($p<0.05$) between condition and sagittal ground reaction forces attributed to load occurred during both movement phases of the "hip knee" and "hip shoulder" task. Additionally, sagittal centre of pressure movement pattern was significantly different under different loading conditions in both movement phases of the "hip knee", "hip shoulder" and "hip twist" tasks.

Movement component of postural control

The trend of muscle activity associated with increased body movement differed according to task demands. Tasks which required increased lever length ("hip shoulder" and "hip reach" tasks) resulted in greater muscle activity in the static condition in comparison to the dynamic condition, resulting in an upward trend in altered muscle activation data with increased load magnitude. Trend of muscle activity attributed to increased trunk movement decreased with increased load in the "hip knee" task. Therefore, dynamic muscle activity was greater than static muscle activity in the "hip knee" task. The "hip twist" task resulted in a combination of the above trends during both movement phases. Muscle activation pattern attributed to increased body movement was significantly different ($p<0.05$) under different external loads in all tasks investigated with the exception of the left rectus abdominis in the second movement phase of the "hip knee" task, the right rectus abdominis during the first movement phase of the "hip shoulder" task and the right latissimus dorsi during the first movement phase of the "hip twist" task. Therefore, increased body movement resulted in significantly different muscle activation patterns under different loading conditions.

Sagittal movement of the centre of pressure attributed to increased trunk motion was significantly different under different loading conditions in both movement phases of the "hip knee" task. The trend of data was the same during both conditions and increased with increased load magnitude. Sagittal movement of the centre of pressure was however linear between the 0.8kg and 1.6kg condition in the second movement phase of the “hip knee” task. Increased body movement resulted in significantly different ground reaction forces in the sagittal, lateral and vertical planes in the “hip shoulder” task. Trend of ground reaction force attributed to increased trunk movement in the “hip shoulder” task was similar in all planes and increased under increased load magnitude. Sagittal movement of the centre of pressure followed the same trend and was significantly different under different loading conditions in both movement phases of the “hip shoulder” task. In both movement phases of the “hip twist” task, ground reaction forces and movement of the centre of pressure attributed to increased trunk movement were significantly altered in all planes and during both movement phases with the exception of sagittal ground reaction forces in the first movement phase. Trend of ground reaction force differed between planes and between movement phases with the exception of vertical ground reaction forces and lateral movement of the centre of pressure which increased under increased loading conditions. Vertical ground reaction forces and sagittal centre of pressure movement were significantly different ($p<0.05$) under different loading conditions in both movement phases of the “hip reach” task. In the second movement phase of the ‘hip reach” task, all ground reaction force variables decreased from the 0.8kg to 1.6kg condition with the exception of the vertical ground reaction force.

A significant interactional effect ($p<0.05$) between condition and data point was found in vertical ground reaction forces and sagittal centre of pressure movement in both movement phases of all tasks investigated. Sagittal ground reaction forces attributed to increased body movement was significantly different in both movement phases of all tasks investigated with the exception of the first movement phase of the ‘hip reach” task.

HYPOTHESES

The null hypothesis regarding the proposed load component of postural control (hypothesis 1) stated that the muscle activity and ground reaction forces attributed to load would be the same with increased load magnitude and as such would be a linear function. Therefore the difference in responses between load one (0.8kg) and load three (4kg) (indicated by A in Figure 32) would be three times the magnitude of the difference between load two (1.6kg) and load three (4kg) (indicated by C in Figure 32) and four times the magnitude of the difference in responses between loads one (0.8kg) and two (1.6kg) (indicated by B on Figure 32). However, it was found that the difference in responses was not the same with increased load magnitude as muscle activity and ground reaction forces attributed to load were significantly different ($p<0.05$) under increased loading conditions. Therefore, the null hypothesis (H_o: A = 4B and A = 3C) was tentatively rejected.

The null hypothesis concerning the proposed movement component of postural control (hypothesis 2) stated that the difference in responses between the static and dynamic condition for all tasks, would be the same when performed under the same external load mass. As muscle activity and ground reaction forces attributed to increased trunk movement were significantly different ($p<0.05$) under different loading condition, the null hypothesis (H_o: x = y = z) was tentatively rejected.

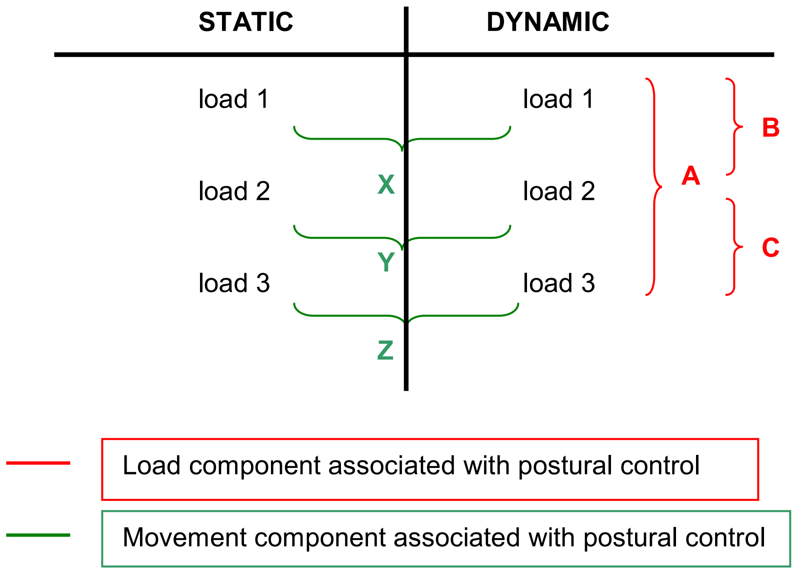


Figure 32: Basic experimental concept.

CONCLUSIONS

Results from the current study emphasise the importance of consideration of the underlying mechanisms of postural control during task completion. Muscle activity and ground reaction forces attributed to load differed significantly under increased loading conditions. Therefore, performing tasks with greater external load resulted in additional strain on the body, in excess of that imposed by task demands. Additionally, muscle activity and ground reaction forces attributed to increased trunk movement was significantly different under different external loads. Consequently, increased trunk movement at higher external loads resulted in additional muscle activity and ground reaction forces in addition to that imposed by task demands. The increased muscle activity associated with load magnitude and trunk motion may serve to increase postural stability during task completion and this additional muscle activity is not accounted for in overall load analyses.

These results may have an impact on the development of future guidelines aimed at reducing the occurrence of low back disorders and low back pain. According to this research, increased external load and increased trunk motion results in additional strain on the body over and above that imposed by task demands. Consequently, in such conditions, guidelines may need to be adapted in order to account for increased muscle activity and body sway.

RECOMMENDATIONS

Future research into the mechanisms of postural control associated with increased external load and increased trunk movement should consider the following recommendations.

Further laboratory investigations are necessary in order to gain a greater understanding of the underlying mechanisms of postural control. Such investigations could include:

a) Analysis of the relationship between muscle activity and ground reaction forces.

b) In order to asses the variability in movement and muscle activation patterns, the coefficient of variation associated with increased load and body movement could be analysed in order to determine the variability of movement under different conditions.

c) Further analysis of the pattern of muscle and movement could be performed to determine the point in the pattern which is most substantially affected by increased load and body movement.

d) Investigations could also further analyse the point at which movement switches from being associated with a decrease in muscle activity (movement efficiency) to an increase in muscle activity (additional strain).

e) Investigations could identify possible differences between responses during the first lift and lower cycle in each task compared to following lift and lower cycles in order to see the effect of learning and subsequent movement modification.

f) Future analysis of the proposed load component of postural control could focus on static tasks in order to analyse the effect of increased load on static tasks.

g) A greater number of tasks could be investigated such that asymmetrical tasks are included to a greater extent. Tasks could also include a greater variety of working heights.

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APPENDICES

APPENDIX A: GENERAL INFORMATION

Letter of information

Pre-test instructions

Informed consent form

LETTER TO THE SUBJECT

Dear ____________________________________

Thank for participating in my Masters thesis entitled “The mechanisms of postural control associated with dynamic lifting tasks”. Your assistance is much appreciated.

The focus of the project is to investigate the muscle activity and ground reaction forces associated with a variety of dynamic and static tasks. The main objective of the study is to determine whether the muscular activity involved in maintaining postural equilibrium is affected by external load and the degree of movement taking place during a task. This study will include both male and female subjects as an increasing amount of females are participating in manual handling jobs.

PROCEDURES

You will be required to attend one testing session of approximately one and a half hours in the Human Kinetics and Ergonomics Department. During the testing session, your stature, mass, arm length and knee height will be recorded. This information will be used to alter the workstation according to your individual anthropometric data. Muscle activity of the trunk muscles will be measured and tests of maximum voluntary contraction will be performed to target each of the muscles tested.

The tasks required for data collection include lifting and lowering a box from a variety of positions around the body. These tasks will be explained to you and you will be allowed to practice each of the tasks. Each of the tasks will take place using two different loads and therefore you are required to perform each of the tasks twice.

Muscle activity of the trunk muscles will be measured using an electromyographic (EMG) unit via electrodes which are placed on the skin. Prior to the placement of the electrodes, the skin surface may need to be cleaned and shaved. After the completion of each of the tasks, you will be required to identify and rate areas of the body experiencing discomfort on a scale of 1 (mild discomfort) to ten (extreme discomfort). This will be explained further during the testing session.

Please note that your data will be kept anonymous and at no time will your name be used. The data collected will be used for statistical purposes and a copy of the data will be kept anonymously in the Human Kinetics and Ergonomics Department and may be used for teaching and research purposes.

Following the completion of data collection, I will gladly discuss the results of the project with you. Thank you for the interest that you have shown and please do not hesitate to contact me in the Human Kinetics and Ergonomics Department if you have any questions.

Yours sincerely

Clare Pettengell
(MSc student – Department of Human Kinetics and Ergonomics)

INSTRUCTION TO SUBJECT PRIOR TO TESTING SESSION

Please refrain from partaking in the following activities **24 hours prior** to testing to allow for all of the conditions to be standardized. Please inform the researcher on the day of the test if you did partake in any of these practices, as this will affect the accuracy of the data obtained:

1. Don’t drink alcohol
2. Try not to take any medication (such as painkillers, flu medications, panado etc)
3. Don’t perform any unaccustomed strenuous exercise

Two hours prior to testing please:

1. Do not consume any stimulants (such as coffee, red bull, coke etc)
2. Please ensure that you eat a good meal 2 hours prior to testing
3. Please wear trainers, shorts and a comfortable t-shirt to the testing session

If you do not adhere to these instructions please inform the researcher on arrival at the testing session.

INFORMED CONSENT FORM

I, ______________________________________, do hereby consent to participate in the study entitled: "THE MECHANISMS OF POSTURAL CONTROL ASSOCIATED WITH DYNAMIC LIFTING TASKS". I agree that I have been fully informed, both verbally and in writing, of the procedures involved in this study. I have also been made aware of any potential risks associated with the protocol including muscle discomfort.

I realize that whilst my anonymity will be protected at all times, my results may be published or used for scientific and statistical purposes. I understand the conditions with which I am expected to comply for the duration of the tests, and any queries I have with regards to this have been answered to my satisfaction.

I will inform the researcher immediately if at any point I experience distress or feel uncomfortable, and am fully aware that I may withdraw from participation in this study at any time.

I have read and understood the above information, as well as the information provided in the letter accompanying this form.

Signed at the Department of Human Kinetics and Ergonomics, Rhodes University, on ____/____/2009.

SUBJECT:______________________ (NAME)______________________ (SIGN)

WITNESS:_______________________(NAME)______________________ (SIGN)

RESEARCHER:___________________(NAME)_____________________ (SIGN)

APPENDIX B: DATA COLLECTION

Equipment checklist
Order of procedures
Anthropometric data sheet
Experimental data sheet

EQUIPMENT CHECKLIST

ADMINISTRATION

- Letter to the subject
- Informed consent form
- Subject data sheet
- Instructions to subjects regarding Body Discomfort

DATA COLLECTION EQUIPMENT

- Toledo scale
- Stadiometer
- Laptop – Megawin programme
  - Bertec force platform analysis software
- Electromyography machine
- Force platform
- Anthropometer
- Body Discomfort scale

OTHER EQUIPMENT

- Milton disinfectant
- Cotton wool
- Razor
- Medical tape

STATIONARY

- Examination pad
- Pencils/Pens
- Long ruler

ORDER OF PROCEDURES

Pre-test: (07h30 each morning)

- Charge batteries
- Switch on force plate to warm up

Testing

1. Welcome
2. Explanation of body discomfort
3. Describe project, protocol and equipment

Biomechanical equipment:

a) The EMG device will be used to measure the activity of the selected muscles. This is done by placing surface electrodes on the skin over the specific muscles. The location of each of the muscles will be found by palpating the muscles or asking to move certain body areas in order to identify the muscle.

b) A force plate is a means of detecting the centre of pressure located beneath the feet. This will provide information regarding of the maintenance of balance throughout the completion of the testing session.

Psychophysical Equipment

At the end of each condition, the body discomfort scale and map will be used and you will be asked to rate a maximum of 3 areas of the body experiencing discomfort on a scale of 1 to 10. One representing little discomfort and 10 representing extreme discomfort.

4. Are there any other questions?
5. Letter to the subject and informed consent form. Please read the letter and attached informed consent form if you are willing to participate.
6. Stature and Mass (Room 30)
7. Anthropometric data
8. Opportunity to practice protocols
9. Any questions?
10. Clean muscle sites with alcohol and shave areas if necessary
11. Locate muscle sites and place electrodes
12. Attach EMG and turn EMG unit on
13. Select EMG protocol and subject on laptop

14. MVC testing – 2 repetitions for 5 seconds (each muscle)
15. Start MVC testing in appropriate order – explain each and show picture
16. Instruct subject as to which condition is being performed first
17. Subject stand in the centre of the force plate feet shoulder width apart – mark foot position
18. Start force plate and EMG
19. Complete condition and after condition, one minute rest
20. Ask body discomfort at the end of each condition
21. After completion of all conditions, seat subject
22. Stop EMG and Force plate
23. Remove EMG and electrodes from subject

End of testing:

- Save data
- Convert EMG data to ASCII files and rename
- Save data to external hard drive, relevant computers and flash sticks
- Clean up for next testing

ANTHROPOMETRIC DATA SHEET

Subject number ________________________________

Name:______________________ Stature (mm):___________________

Age:_______________________ Hand dominance:_______________

Body mass (kg):_____________

Anthropometric data

Variable	**Measurement (mm)**
Standing elbow height	
Standing knee height	
Hip Height	
Arm length	

EXPERIMENTATION DATA SHEET

SUBJECT NO.:________________

EMG:

START TIME: ____________________ END TIME:____________________

FORCE PLATE:

START TIME:____________________ END TIME:______________________

CONDITION	START	FINNISH	BODY DISCOMFORT AREA (RATING)	COMMENTS

CONDITION	START	FINNISH	DISCOMFORT AREA (RATING)	COMMENTS

APPENDIX C: PATTERN OF EMG AND GROUND REACTION FORCES

"Hip knee" task

"Hip shoulder" task

"Hip twist" task

"Hip reach" task

ELECTROMYOGRAPHY

HIP KNEE TASK

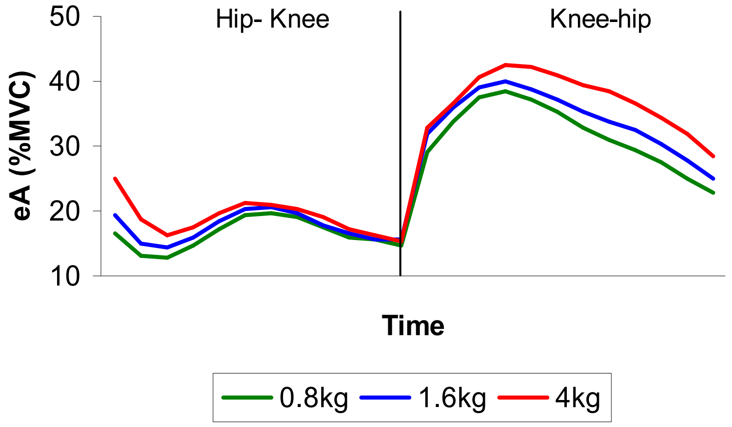


Figure 33: Muscle activation pattern of the left lumbar erector spine muscle.

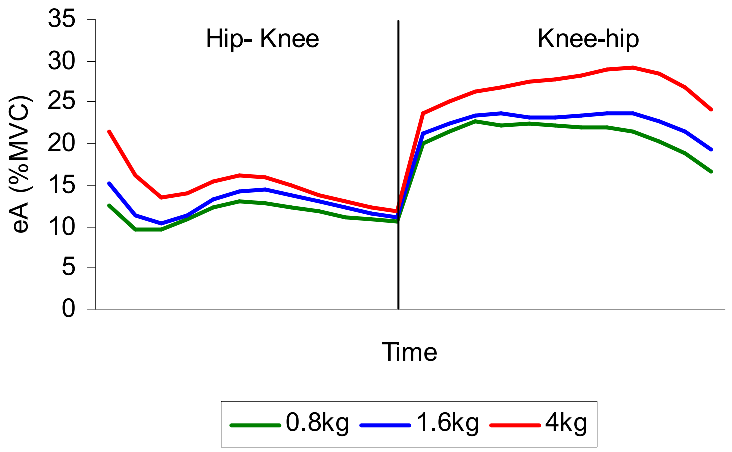


Figure 34: Muscle activation pattern of the left thoracic erector spinae muscle.

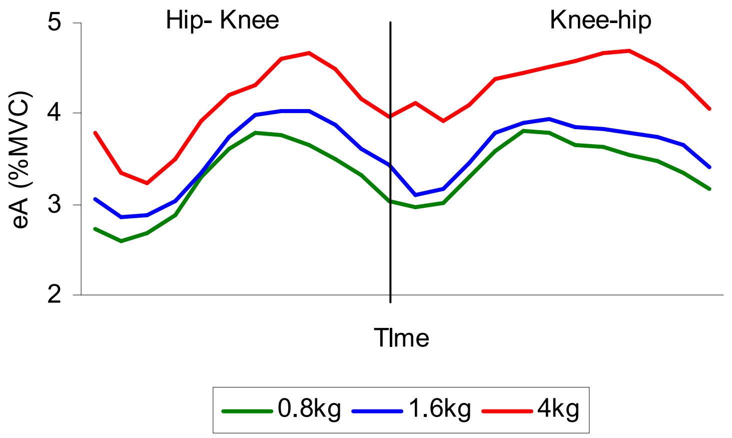


Figure 35: Muscle activation pattern of the left external oblique muscle during the "hip knee" task.

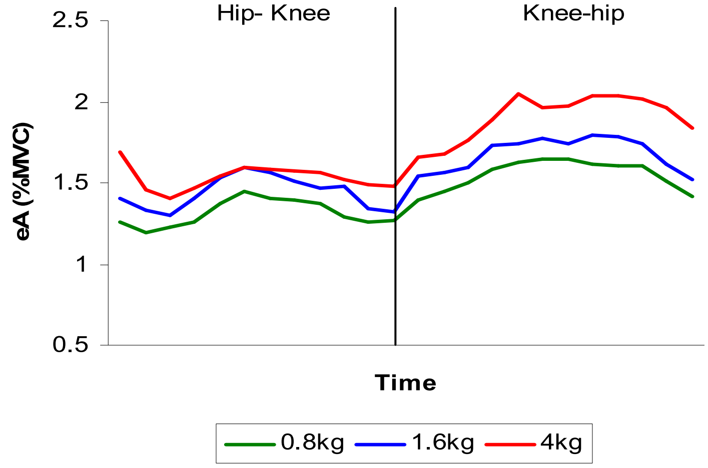


Figure 36: Muscle activation pattern of the left rectus abdominis muscle during the "hip knee" task.

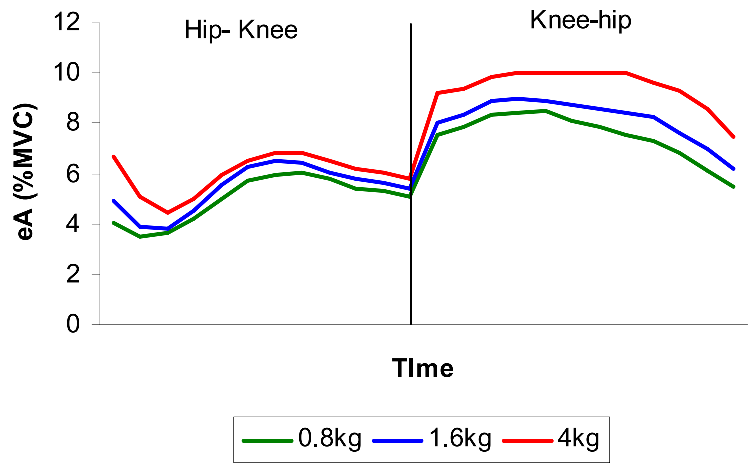


Figure 37: Muscle activation pattern of the left internal oblique muscle during the "hip knee" task.

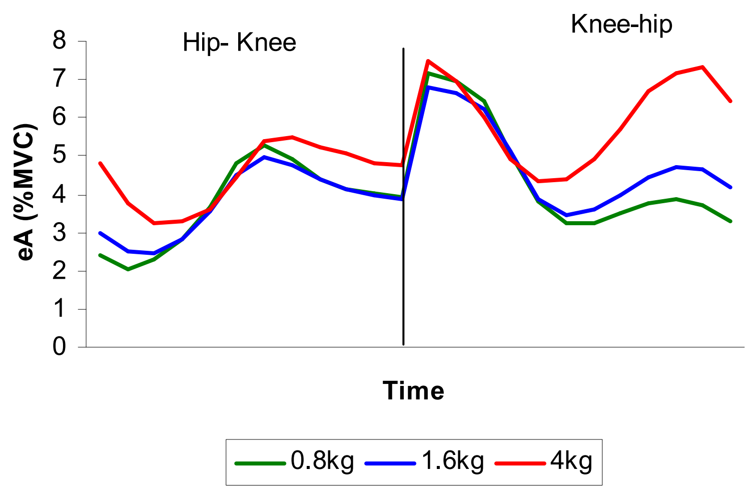
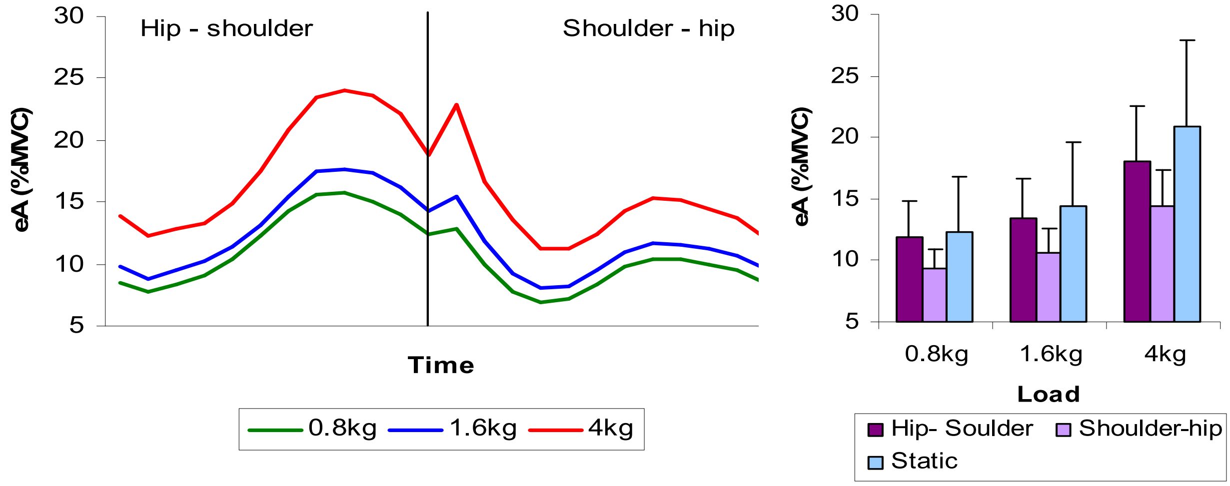


Figure 38: Muscle activation pattern of the left latissimus dorsi muscle during the "hip knee" task.

HIP SHOULDER TASK



(A) (B)

Figure 39: Muscle activation pattern (A) and average activation (B) of the right lumbar erector spinae muscle during the hip shoulder task.

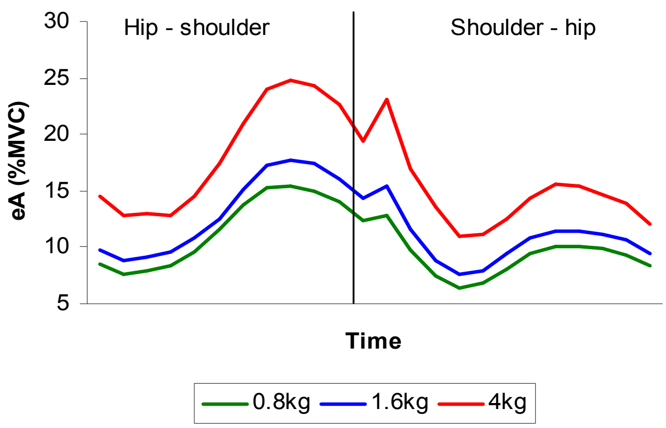
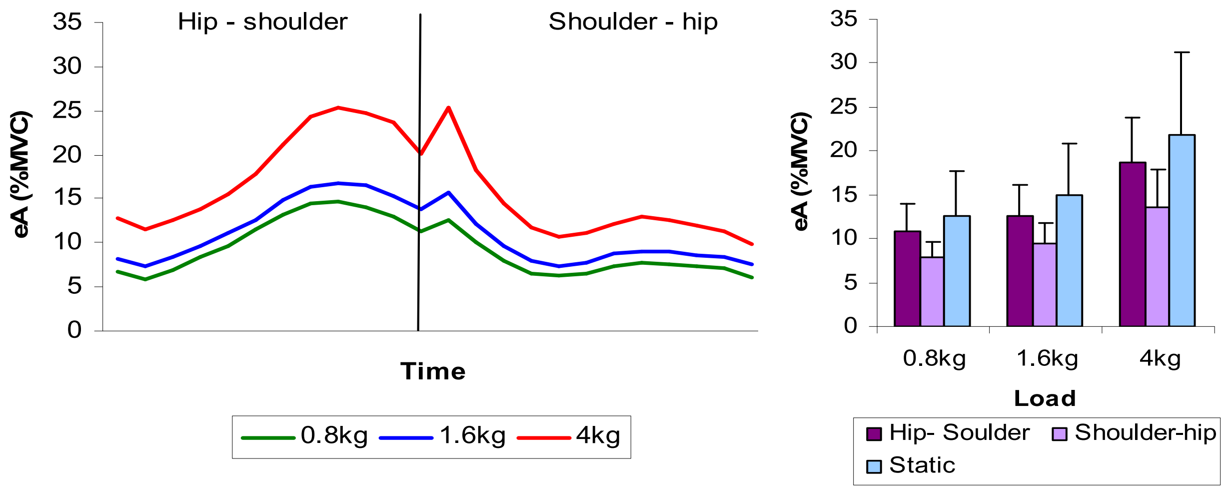


Figure 40: Muscle activation pattern of the left lumbar erector spinae muscle.



(A) (B)

Figure 41: Muscle activation pattern (A) and average activation (B) of the right thoracic erector spinae muscle during the “hip shoulder” task.

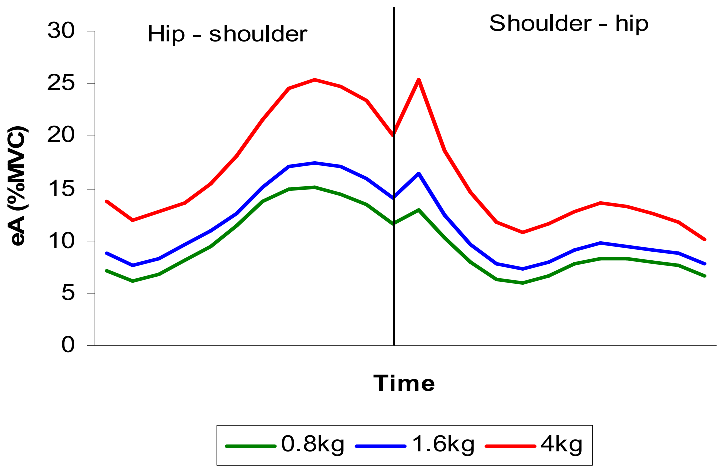


Figure 42: Muscle activation pattern of the left thoracic erector spinae.

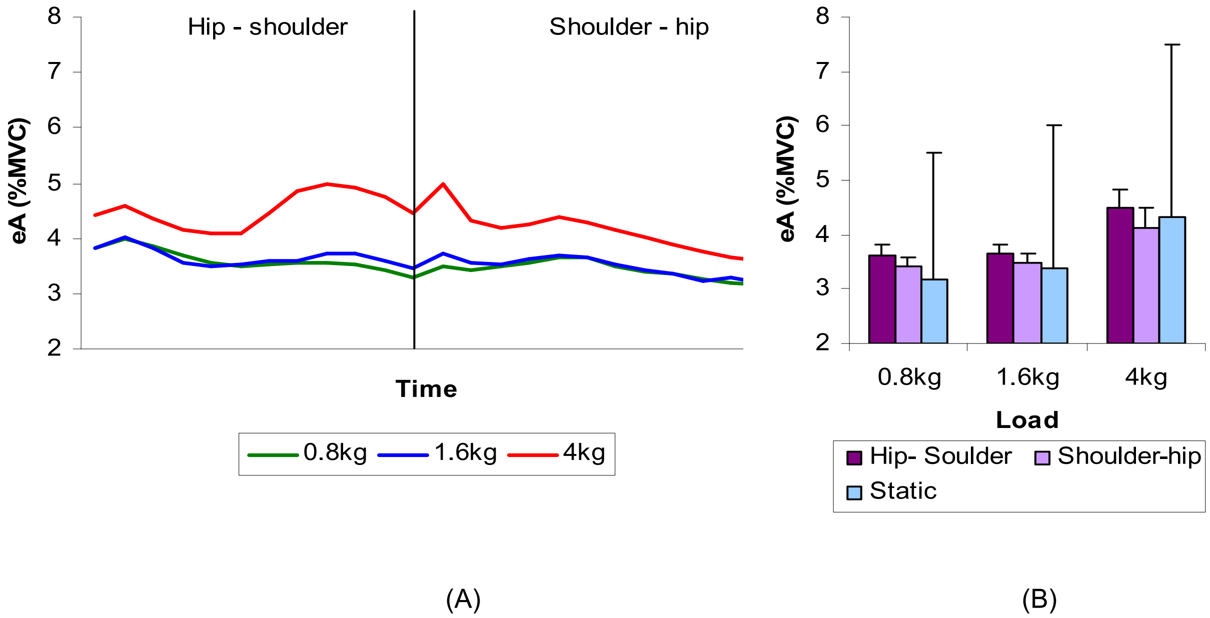


Figure 43: Muscle activation pattern (A) and average activation (B) of the *right external oblique* muscle during the "hip shoulder" task.

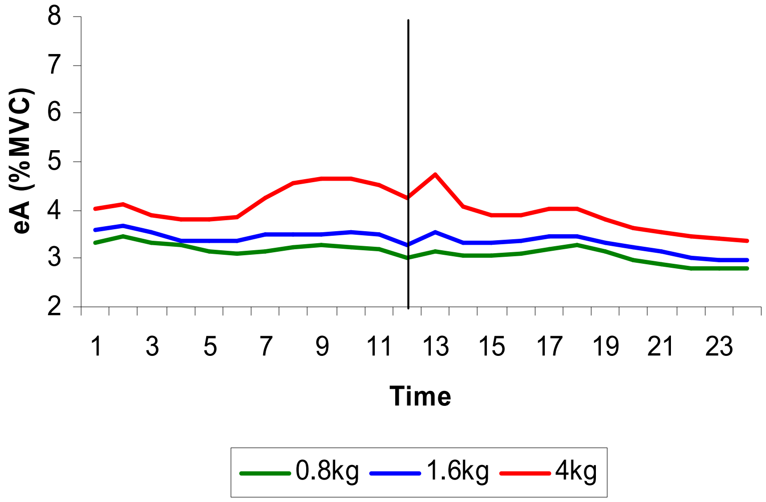
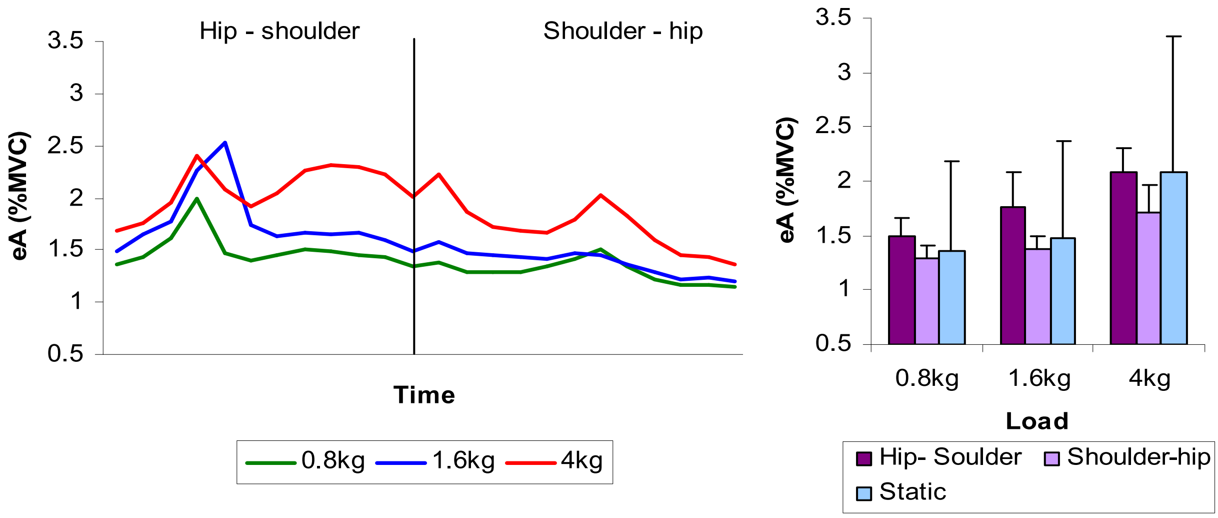
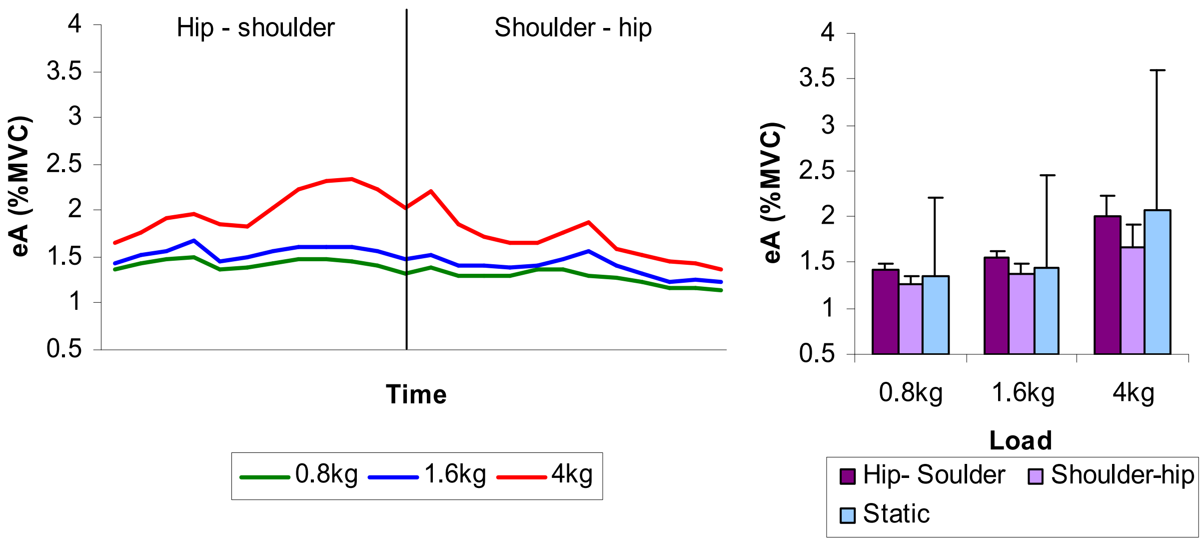


Figure 44: Muscle activation pattern of left external oblique during the "hip shoulder" task.



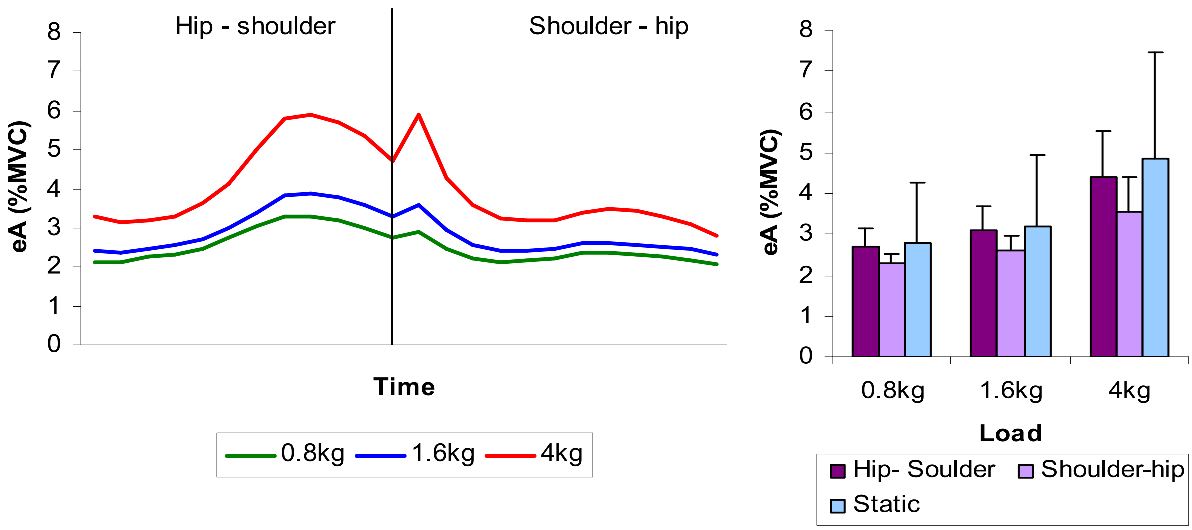
(A) (B)

Figure 45: Muscle activation pattern (A) and average activation (B) of the right rectus abdominis muscle during the "hip shoulder" task.



(A) (B)

Figure 46: Muscle activation pattern (A) and average activation (B) of the left rectus abdominis muscle during the "hip shoulder" task.



(A) (B)

Figure 47: Muscle activation pattern (A) and average activation (B) of the right internal oblique muscle during the "hip shoulder" task.

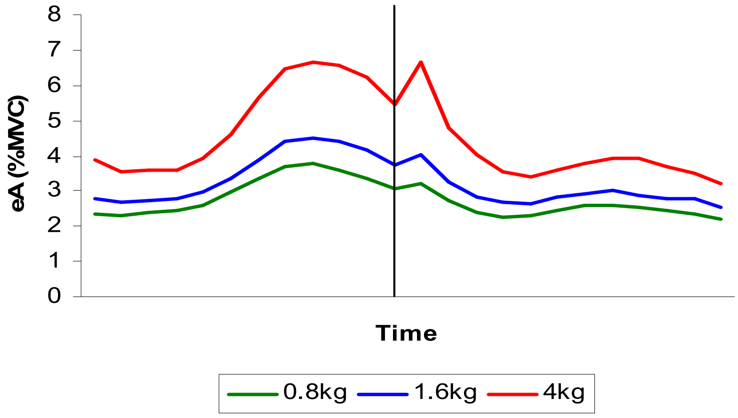


Figure 48: Muscle activation pattern of the left internal oblique during the "hip shoulder" task.

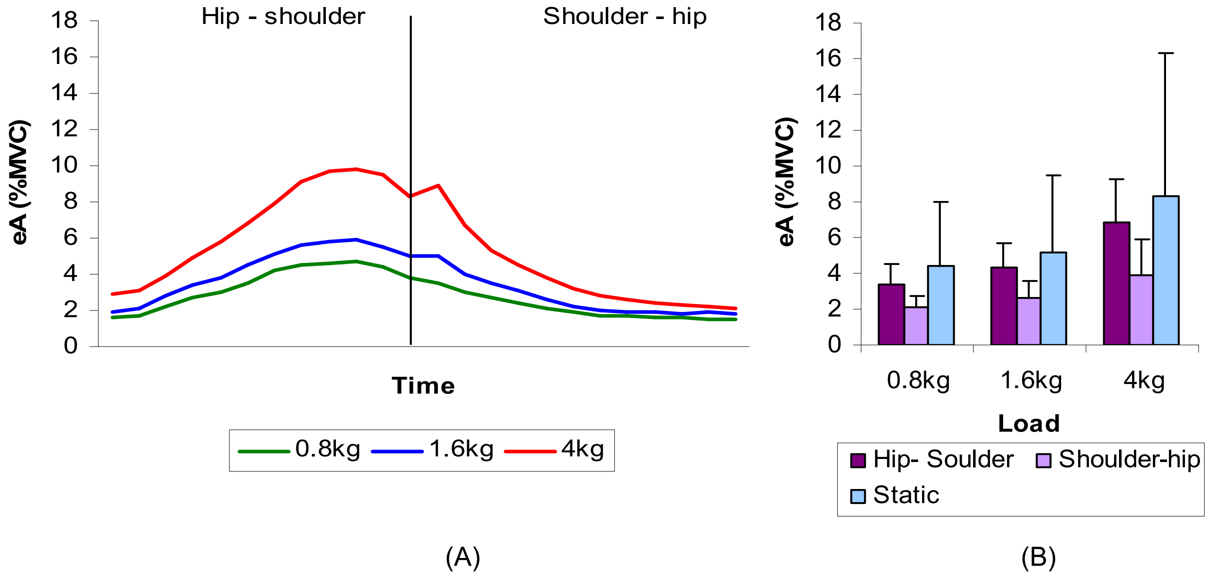


Figure 49: Muscle activation pattern (A) and average activation (B) of the right latissimus dorsi muscle during the "hip shoulder" task.

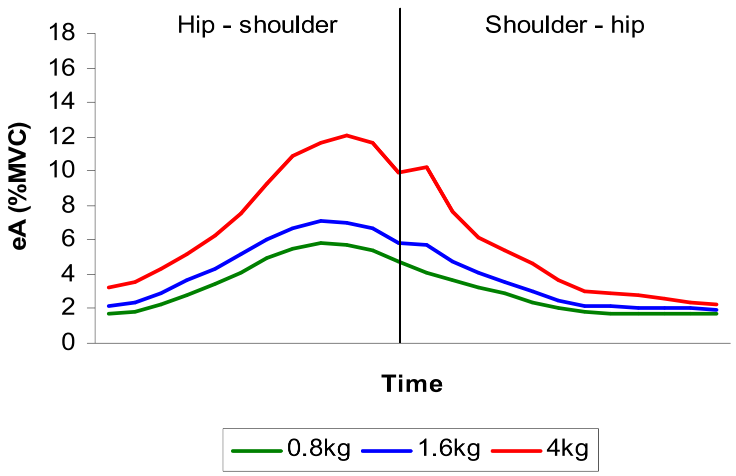
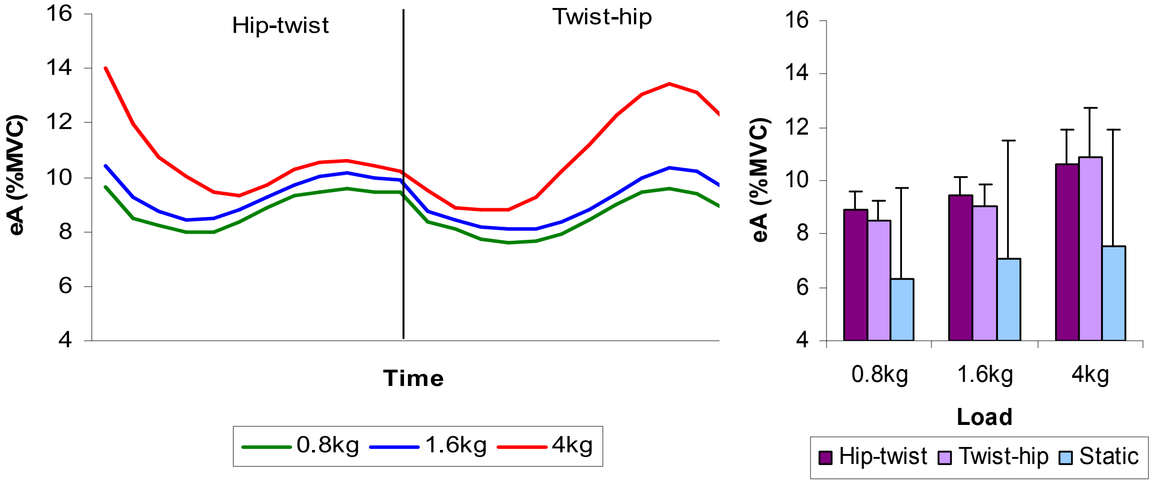


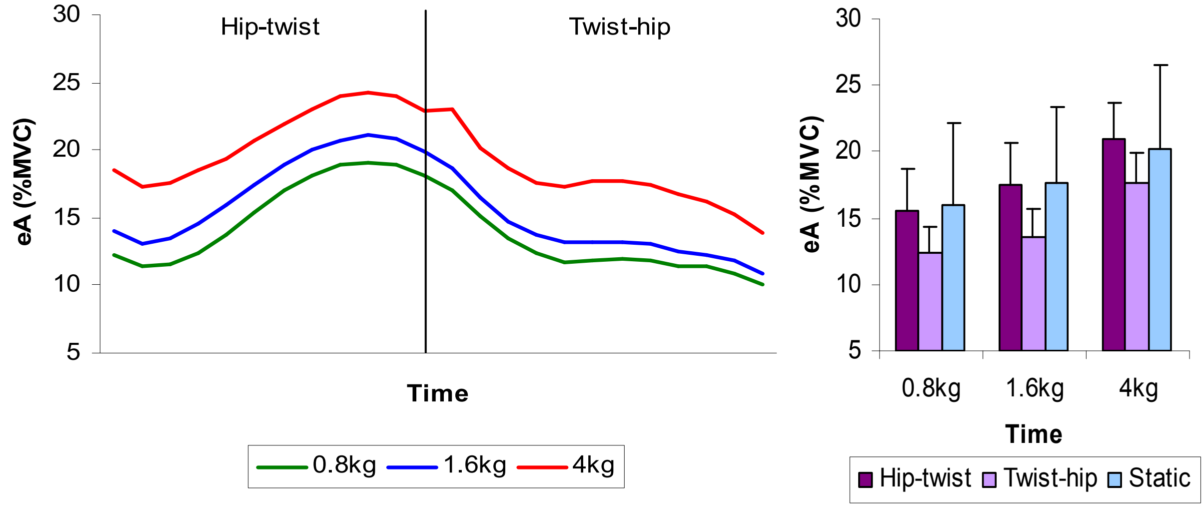
Figure 50: Muscle activation pattern of the left latissimus dorsi muscle during the "hip shoulder" task.

HIP TWIST TASK



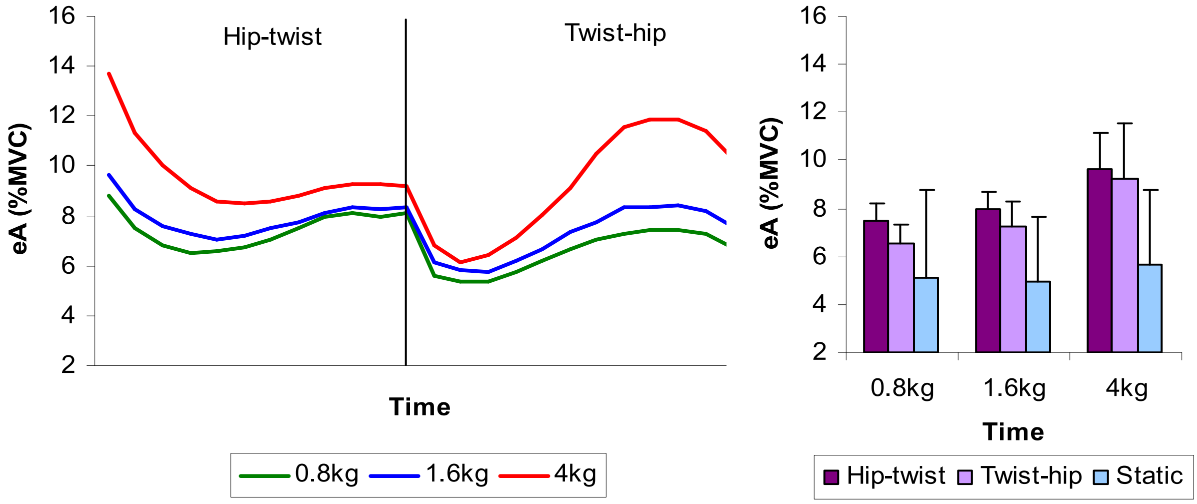
(A) (B)

Figure 51: Muscle activation pattern (A) and average activation (B) of the right lumbar erector spinae muscle during the "hip twist" task.



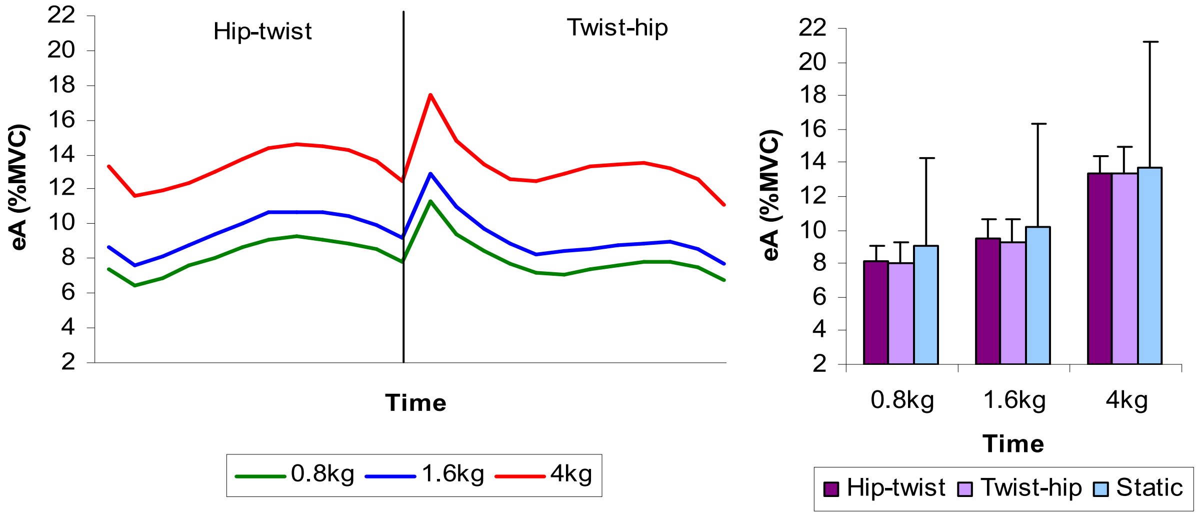
(A) (B)

Figure 52: Muscle activation pattern (A) and average activation (B) of the left lumbar erector spinae muscle during the "hip twist" task.



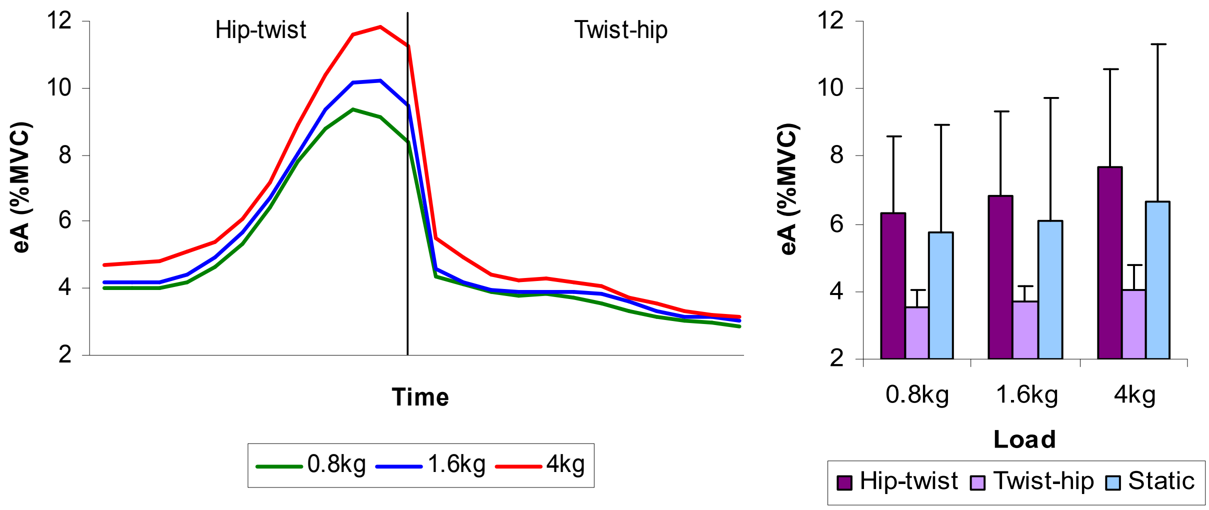
(A) (B)

Figure 53: Muscle activation pattern (A) and average activation (B) of the right thoracic erector spinae muscle during the "hip twist" task.



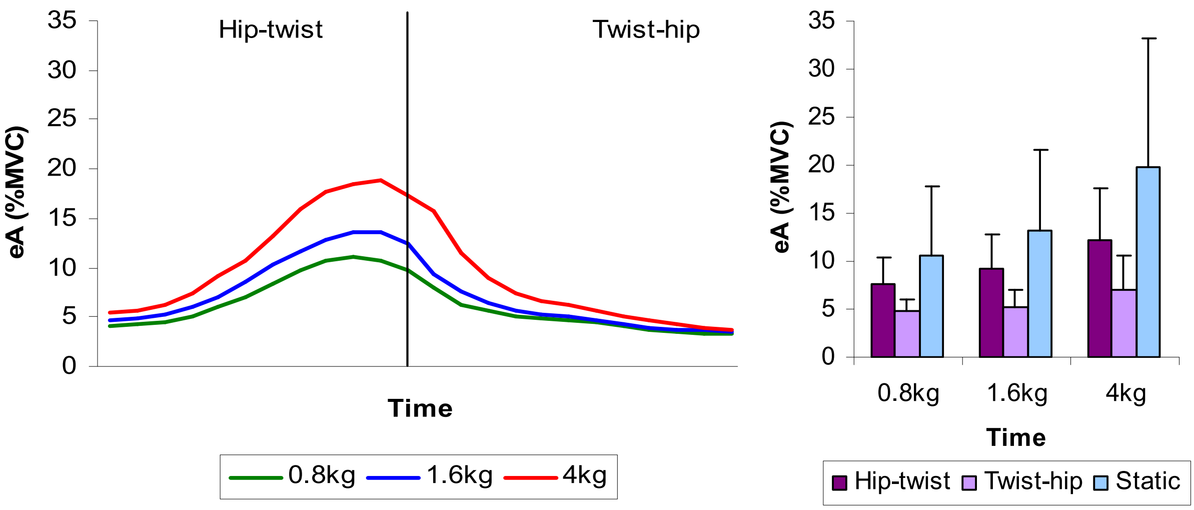
(A) (B)

Figure 54: Muscle activation pattern (A) and average activation (B) of the left thoracic erector spinae muscle during the "hip twist" task.



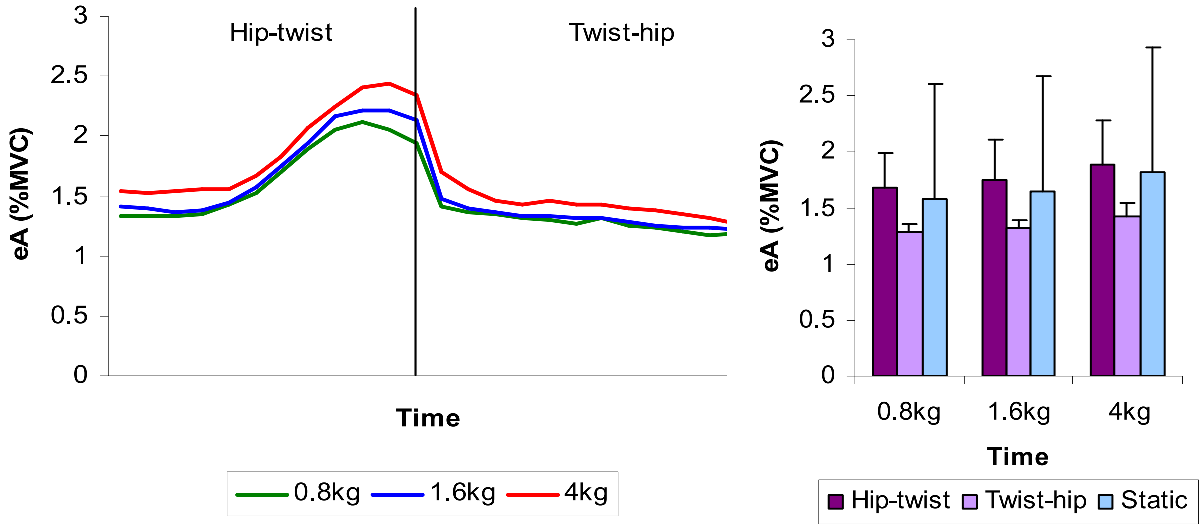
(A) (B)

Figure 55: Muscle activation pattern (A) and average activation (B) of the right external oblique muscle during the "hip twist" task.



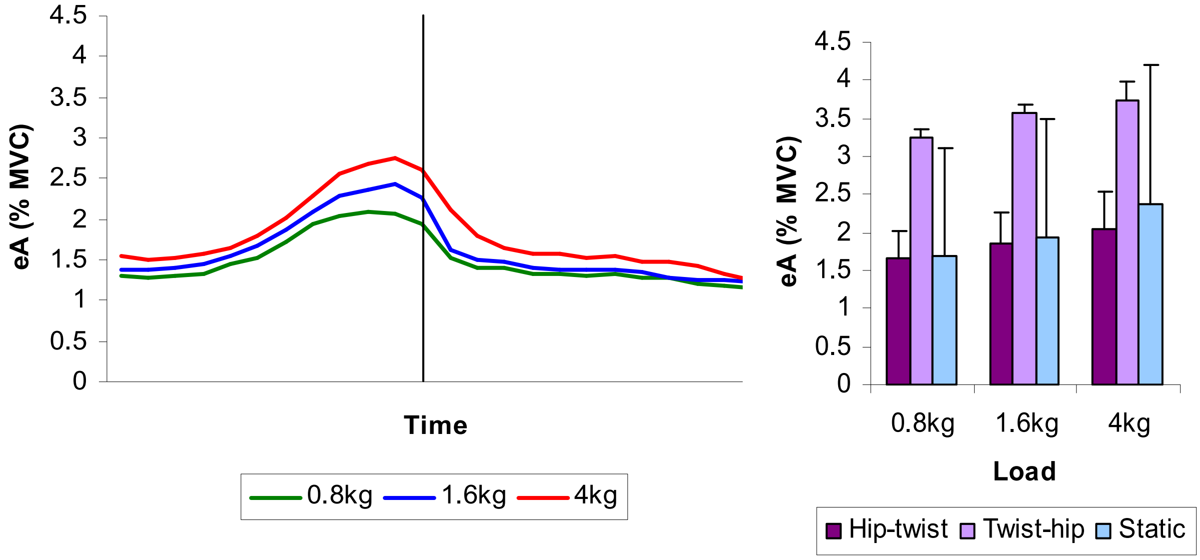
(A) (B)

Figure 56: Muscle activation pattern (A) and average activation (B) of the left external oblique muscle during the "hip twist" task.



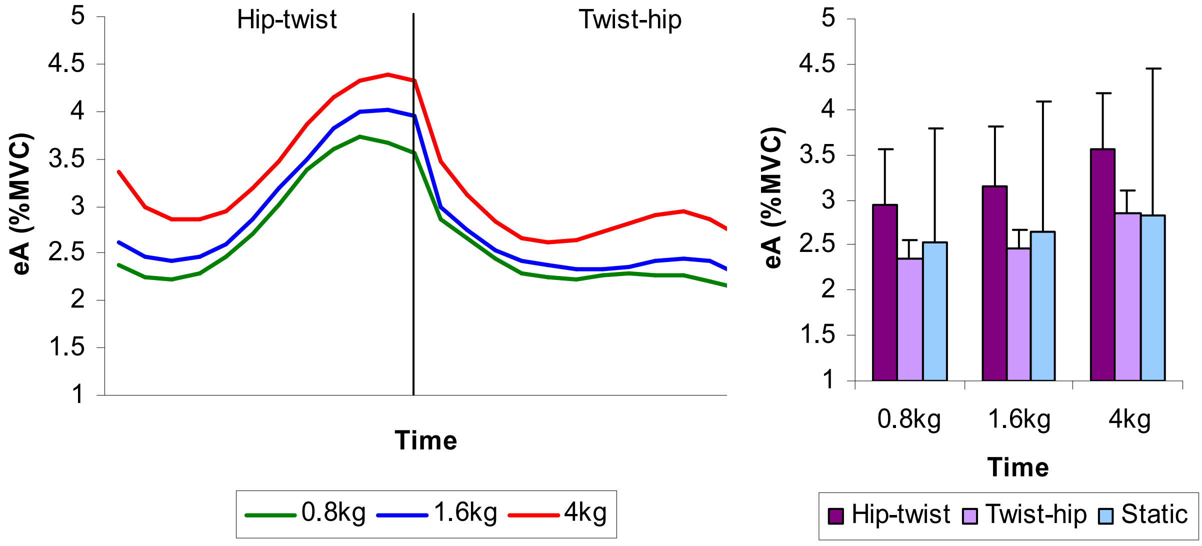
(A) (B)

Figure 57: Muscle activation pattern (A) and average activation (B) of the right rectus abdominis muscle during the "hip twist" task.



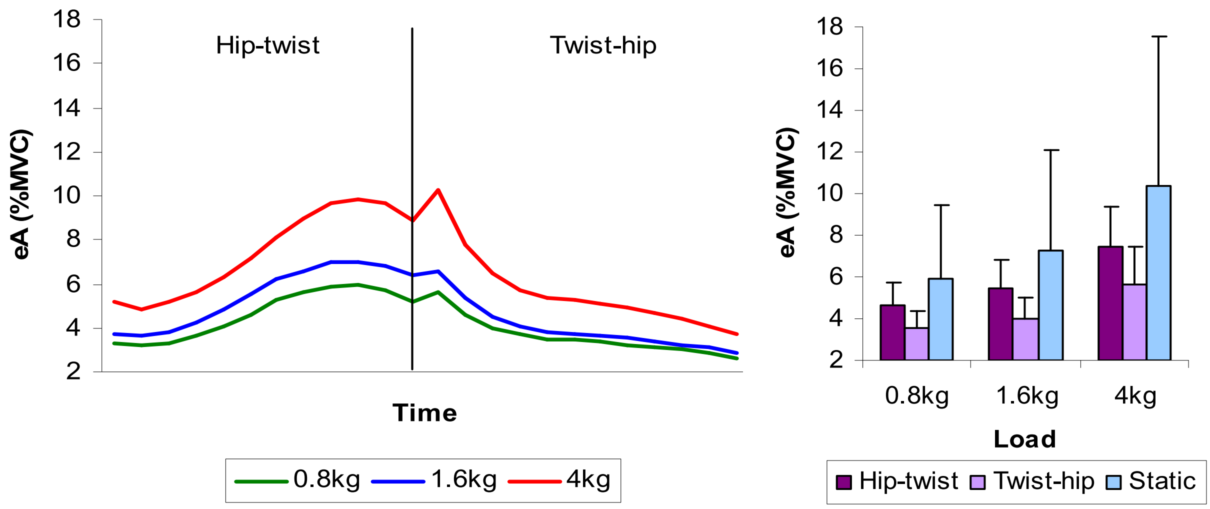
(A) **(B)**

Figure 58: Muscle activation pattern (A) and average activation (B) of the left rectus abdominis during the "hip twist" task.



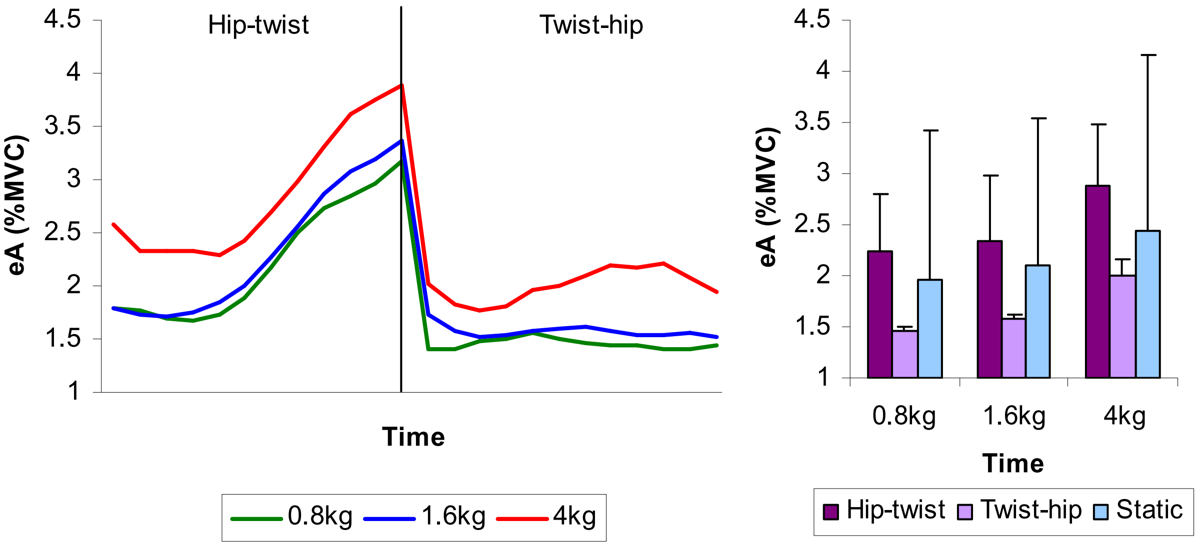
(A) (B)

Figure 59: Muscle activation pattern (A) and average activation (B) of the right internal oblique muscle during the "hip twist" task.



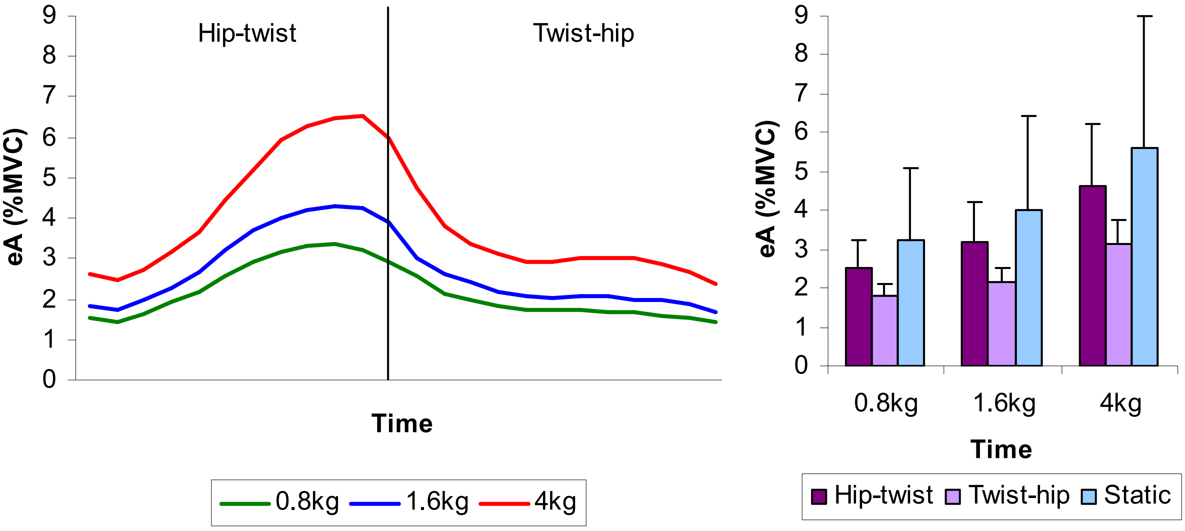
(A) (B)

Figure 60: Muscle activation pattern (A) and average activation (B) of the left internal oblique muscle during the "hip twist" task.



(A) (B)

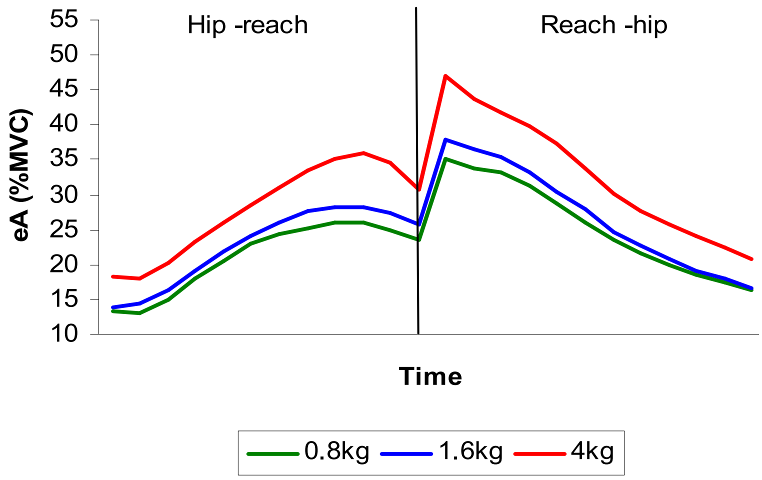
Figure 61: Muscle activation pattern (A) and average activation (B) of the right latissimus dorsi muscle during the "hip twist" task.



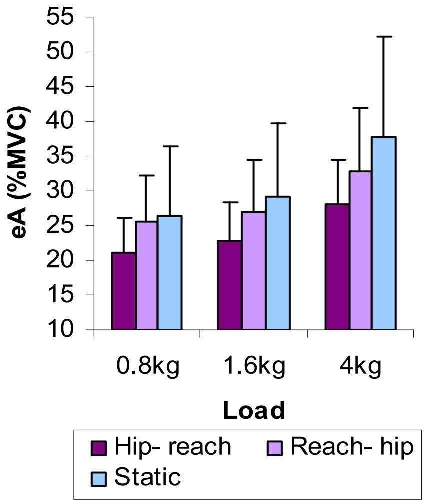
(A) (B)

Figure 62: Muscle activation pattern (A) and average activation (B) of the left latissimus dorsi muscle during the "hip twist" task.

HIP REACH TASK



(A)



(B)

Figure 63: Muscle activation pattern (A) and average activation (B) of the right lumbar erector spinae muscle during the "hip reach" task.

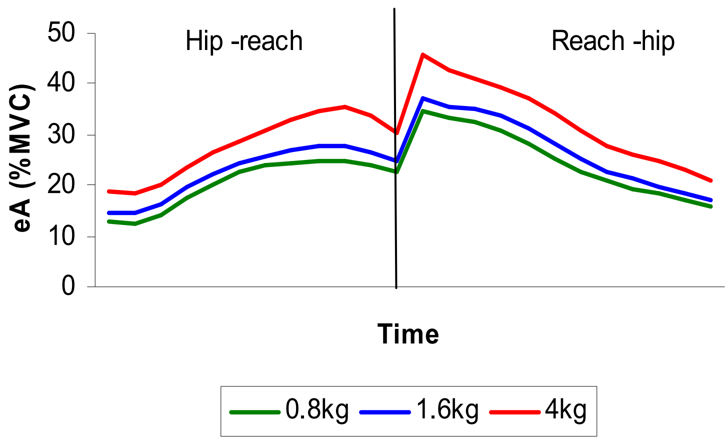
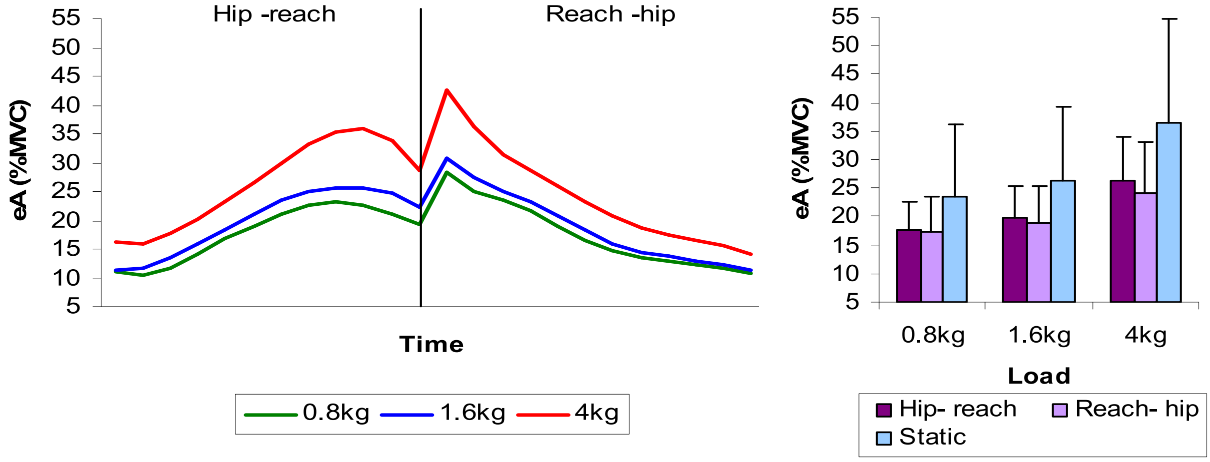


Figure 64: Muscle activation pattern of the left lumbar erector spinae during the "hip reach" task.



(A) (B)

Figure 65: Muscle activation pattern (A) and average activation (B) of the right thoracic erector spinae muscle during the "hip reach" task.

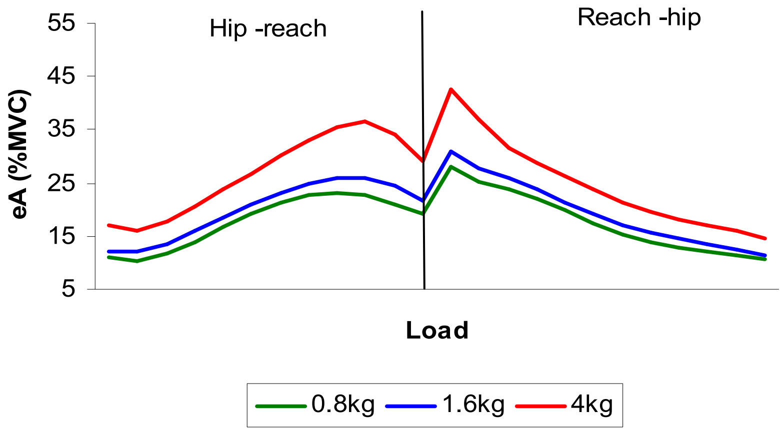
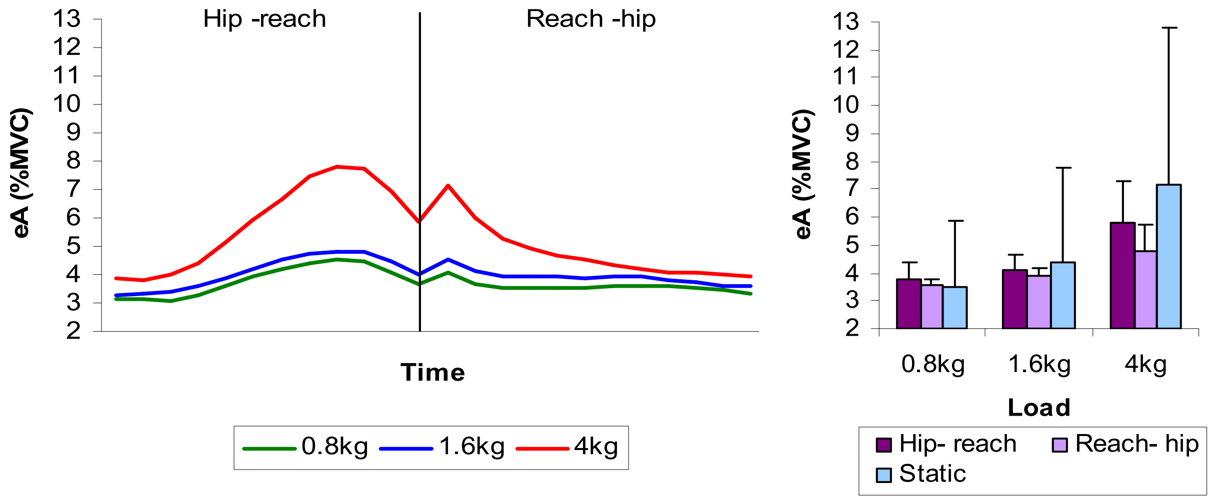


Figure 66: Muscle activation pattern of the left thoracic erector spinae muscle during the "hip reach" task.



(A) (B)

Figure 67: Muscle activation pattern (A) and average activation (B) of the right external oblique muscle during the "hip reach" task.

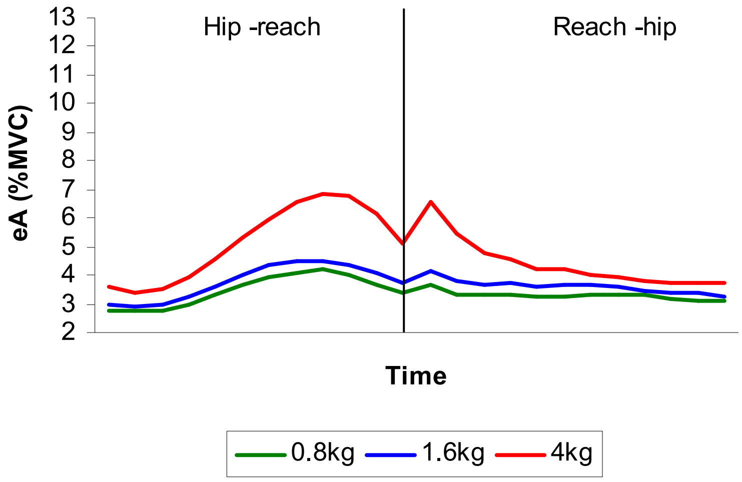
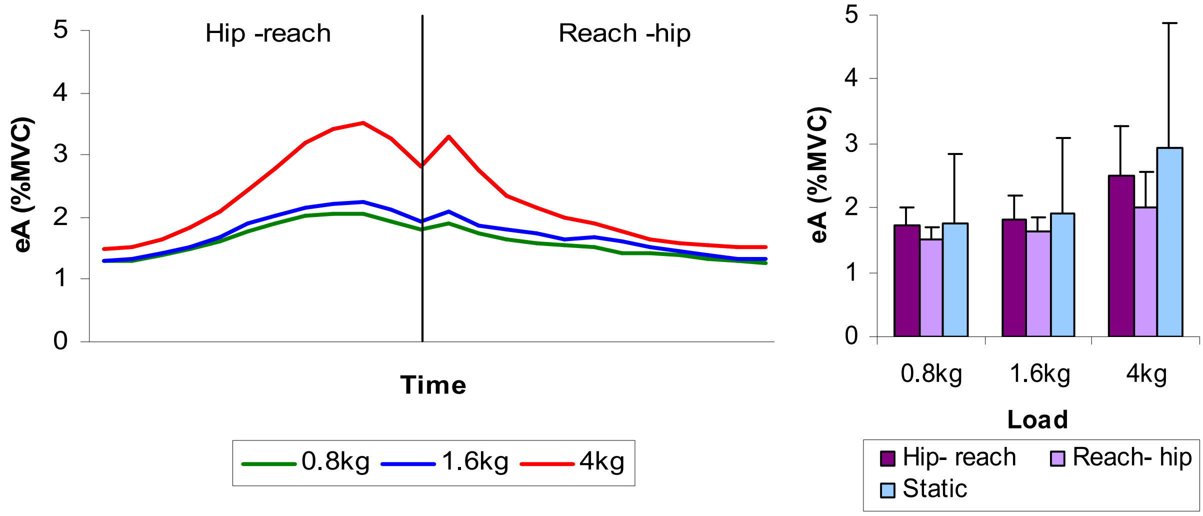


Figure 68: Muscle activation pattern of the left external oblique during the "hip reach" task.



(A) (B)

Figure 69: Muscle activation pattern (A) and average activation (B) of the right rectus abdominis muscle during the "hip reach" task.

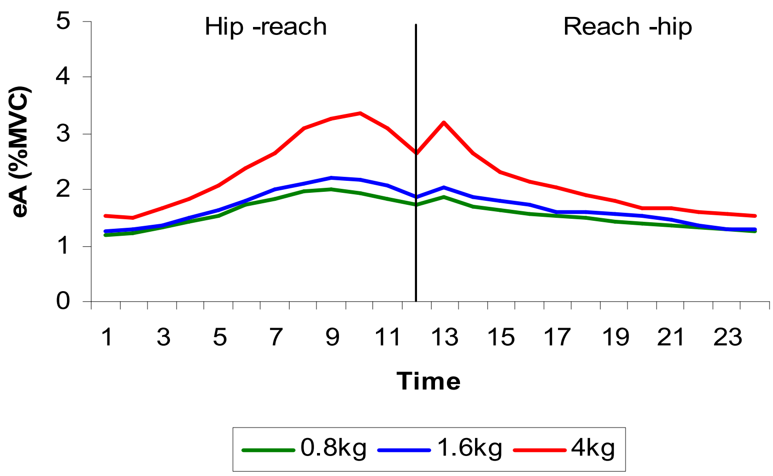
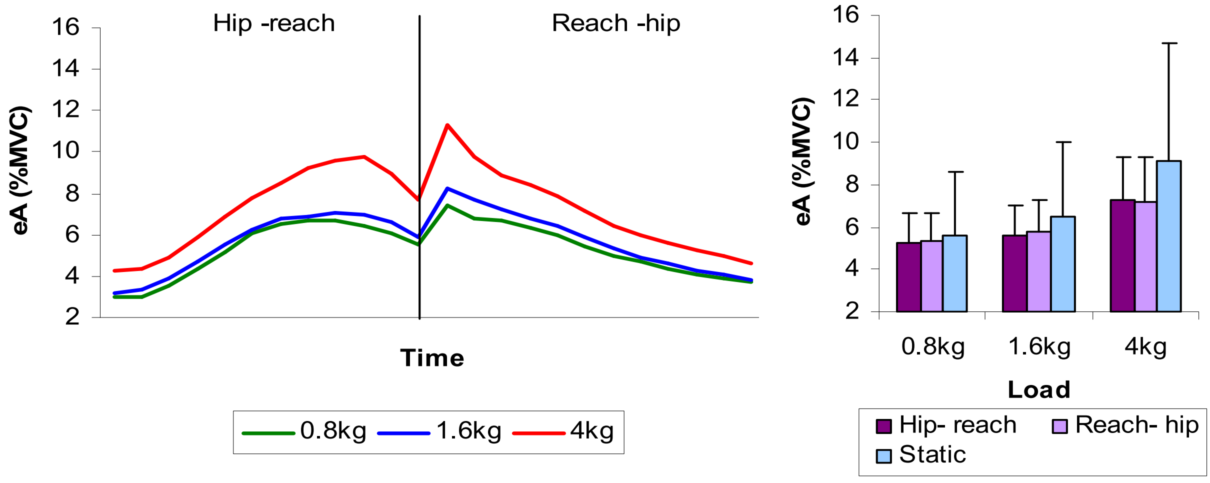


Figure 70: Muscle activation pattern of the left rectus abdominis muscle in the "hip reach" task.



(A) (B)

Figure 71: Muscle activation pattern (A) and average activation (B) of the right internal oblique muscle during the “hip reach” task.

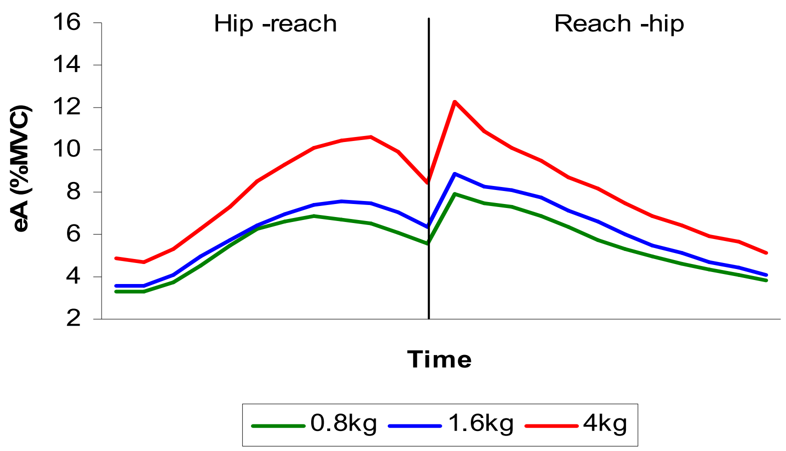
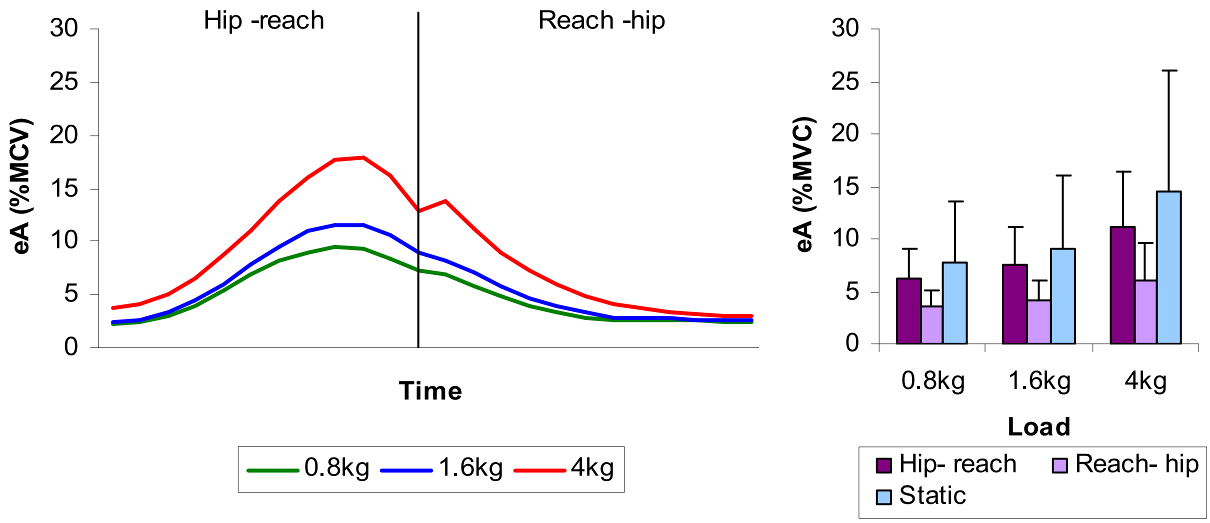


Figure 72: Muscle activation pattern of the left internal oblique during the “hip reach” task.



(A) (B)

Figure 73: Muscle activation pattern (A) and average activation (B) of the right latissimus dorsi muscle during the “hip reach” task.

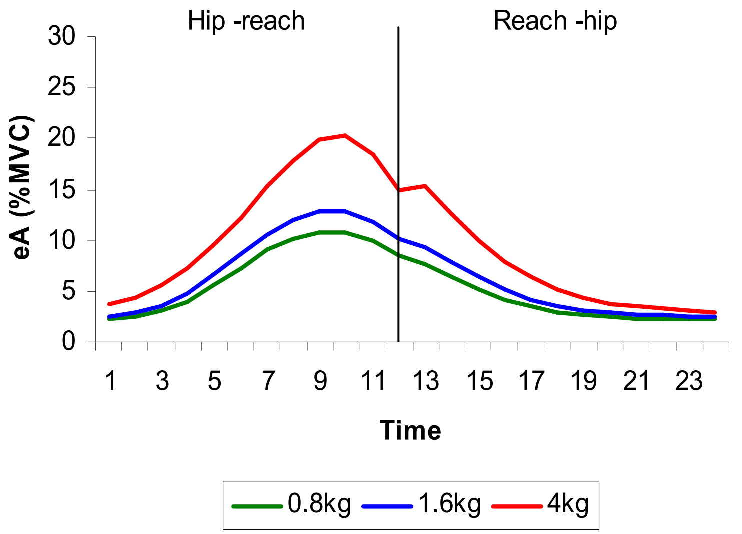
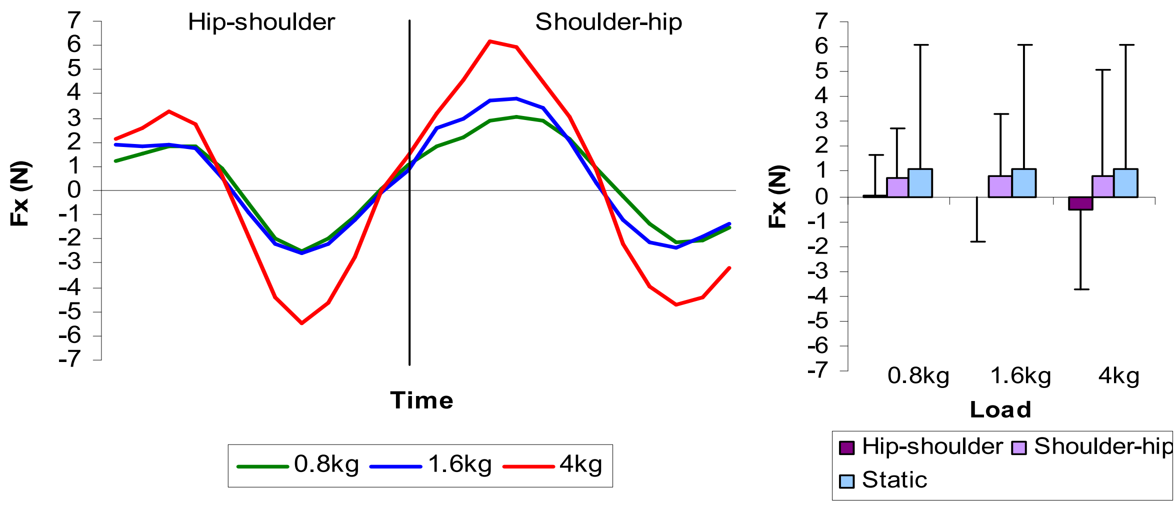


Figure 74: Muscle activation pattern of the left latissimus dorsi muscle during the "hip reach" task.

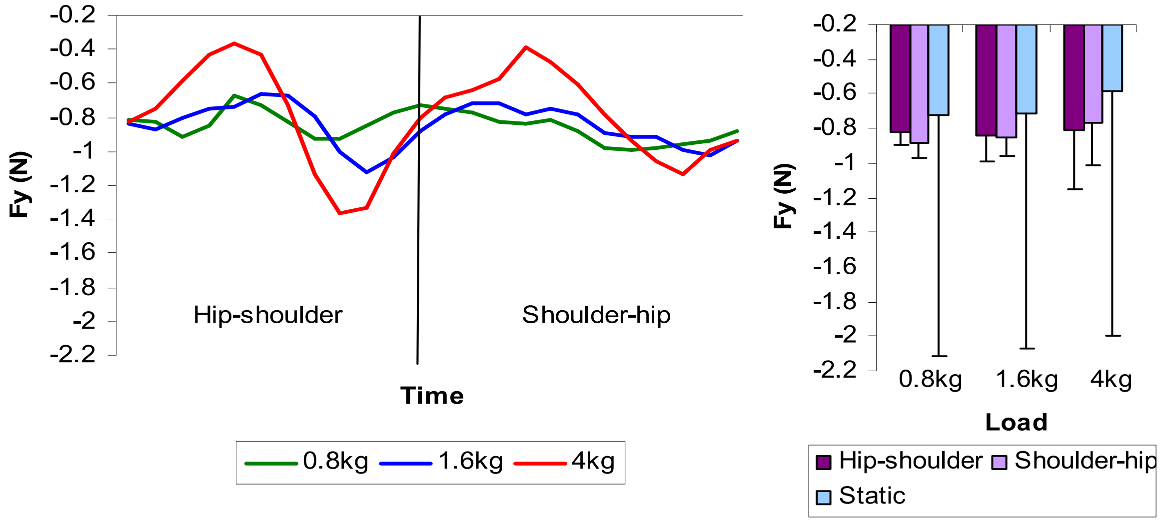
GROUND REACTION FORCES

HIP SHOULDER TASK



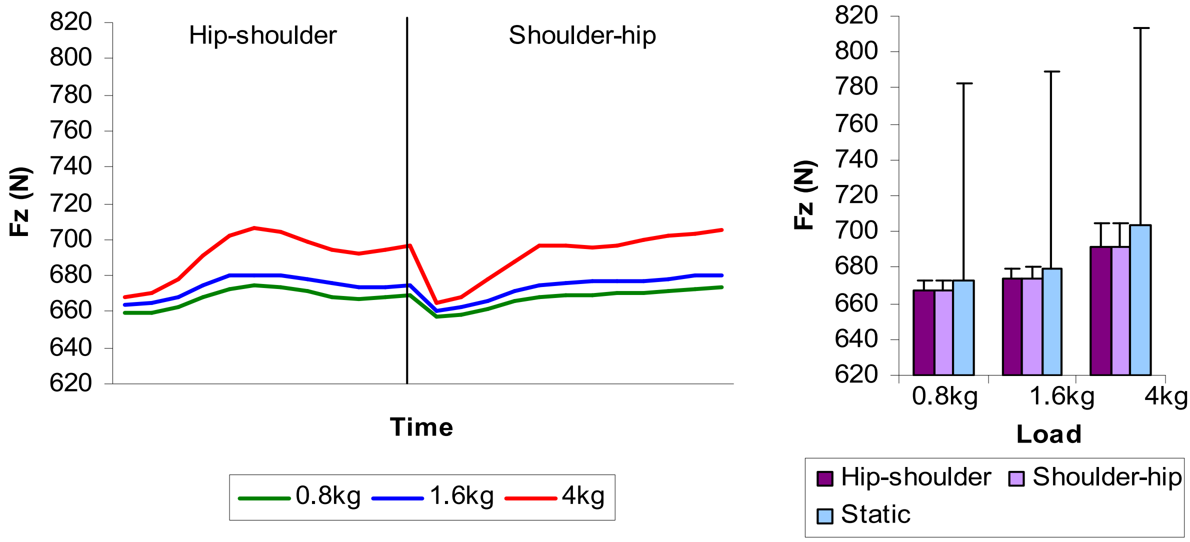
(A) **(B)**

Figure 75: Pattern (A) and average (B) sagittal ground reaction forces under each loading condition during the "hip shoulder" task.



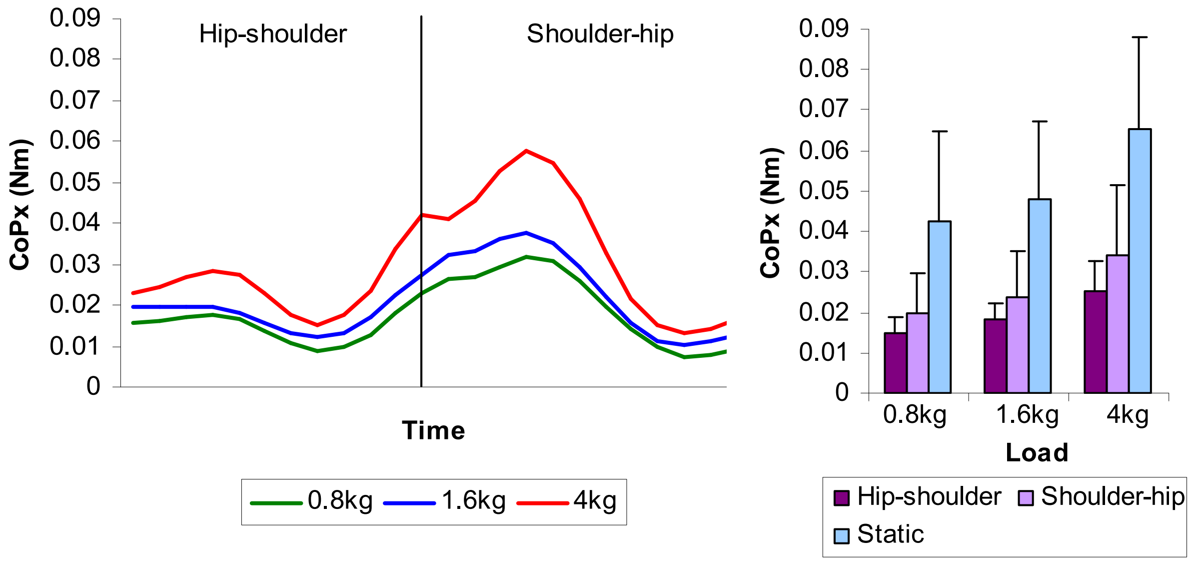
(A) **(B)**

Figure 76: Pattern (A) and average (B) lateral ground reaction forces under each loading condition during the "hip shoulder" task.



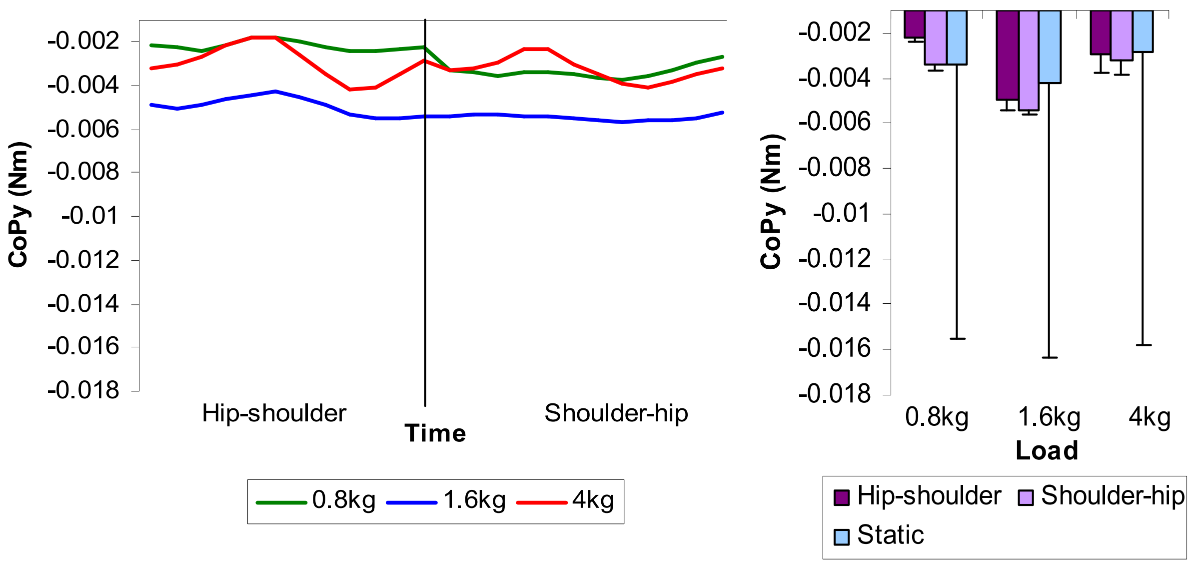
(A) **(B)**

Figure 77: Pattern (A) and average (B) vertical ground reaction forces under each loading condition during the "hip shoulder" task.



(A) (B)

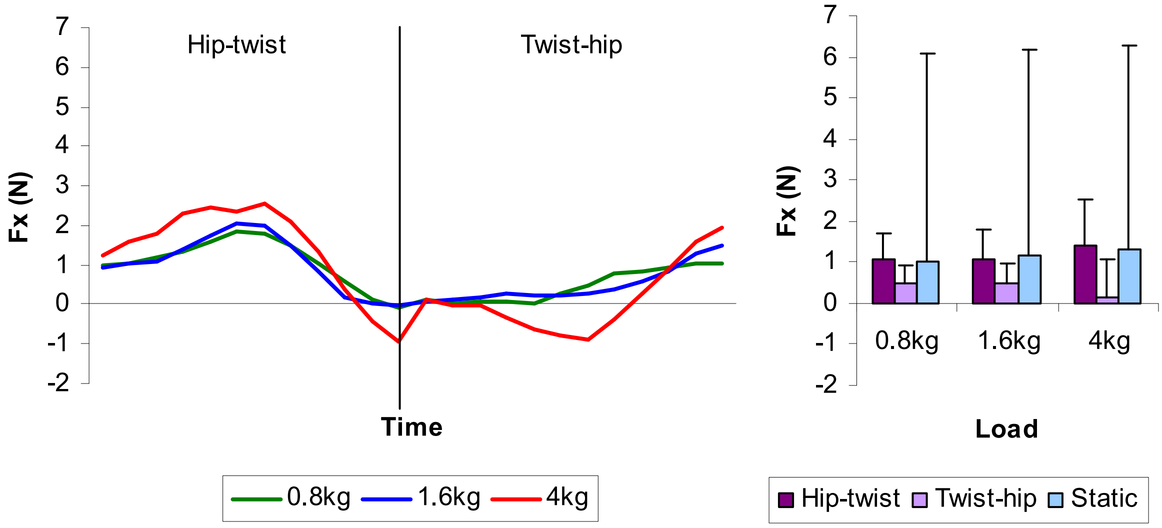
Figure 78: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the anterior/posterior direction under each loading condition during the "hip shoulder" task.



(A) (B)

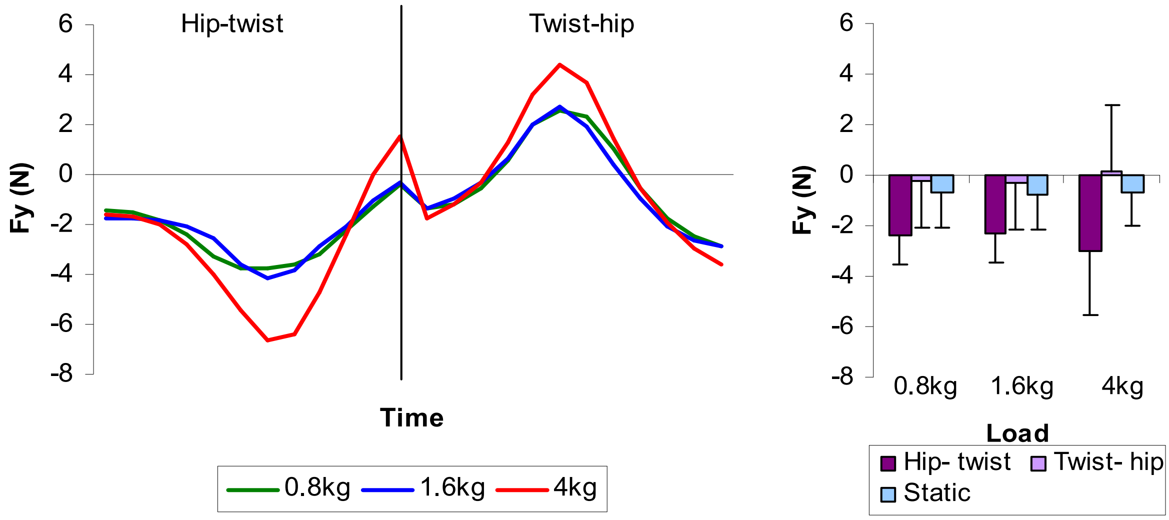
Figure 79: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the lateral direction under each loading condition during the "hip shoulder" task.

HIP TWIST TASK



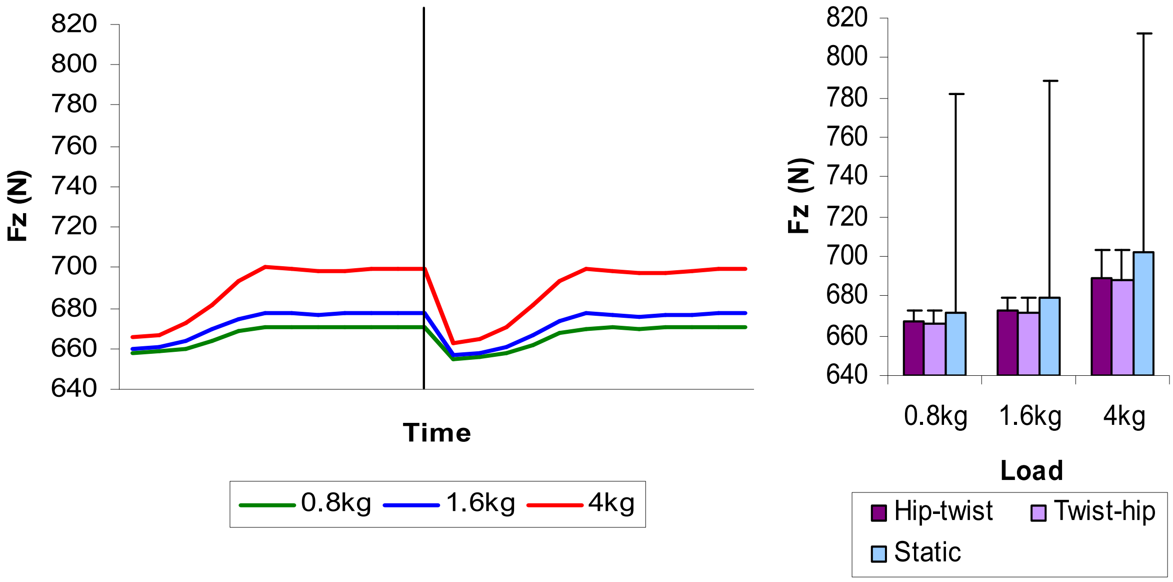
(A) **(B)**

Figure 80: Pattern (A) and average (B) sagittal ground reaction forces of each loading condition during the "hip twist" task.



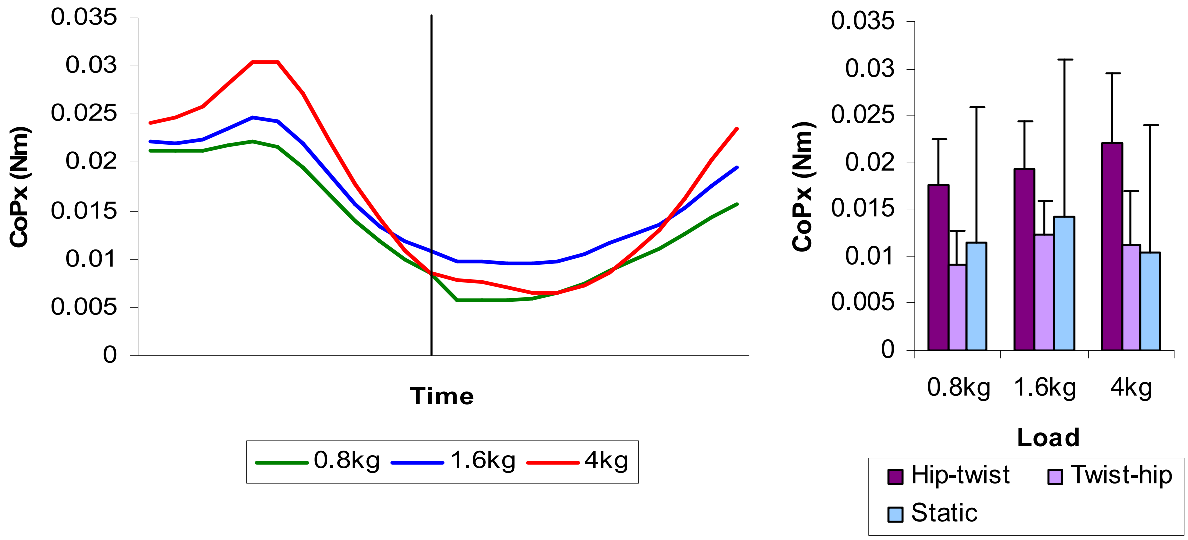
(A) **(B)**

Figure 81: Pattern (A) and average (B) lateral ground reaction forces of each loading condition during the "hip twist" task.



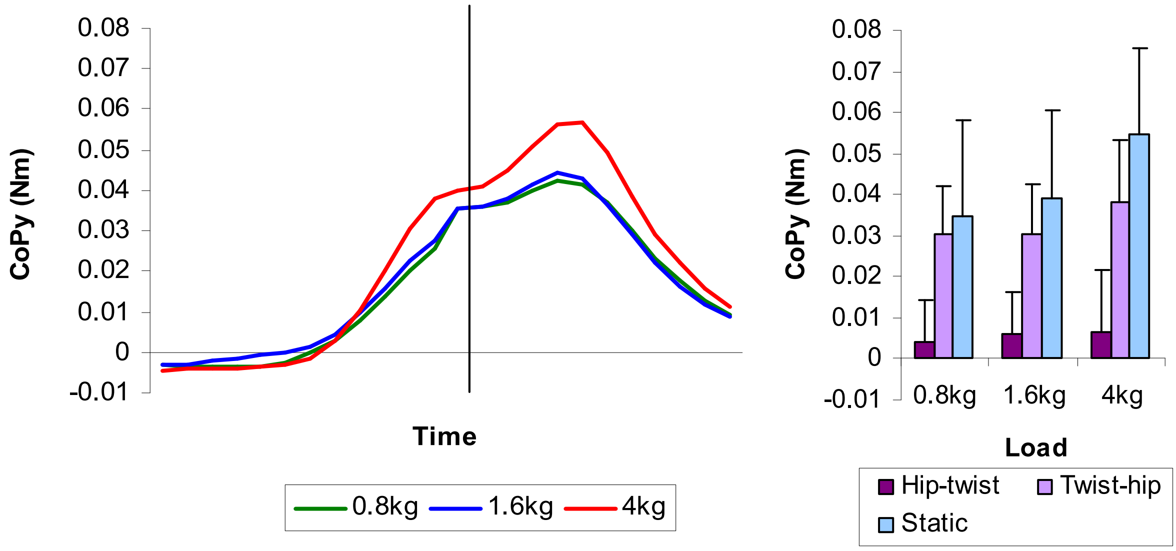
(A) (B)

Figure 82: Pattern (A) and average (B) vertical ground reaction forces of each loading condition during the "hip twist" task.



(A) (B)

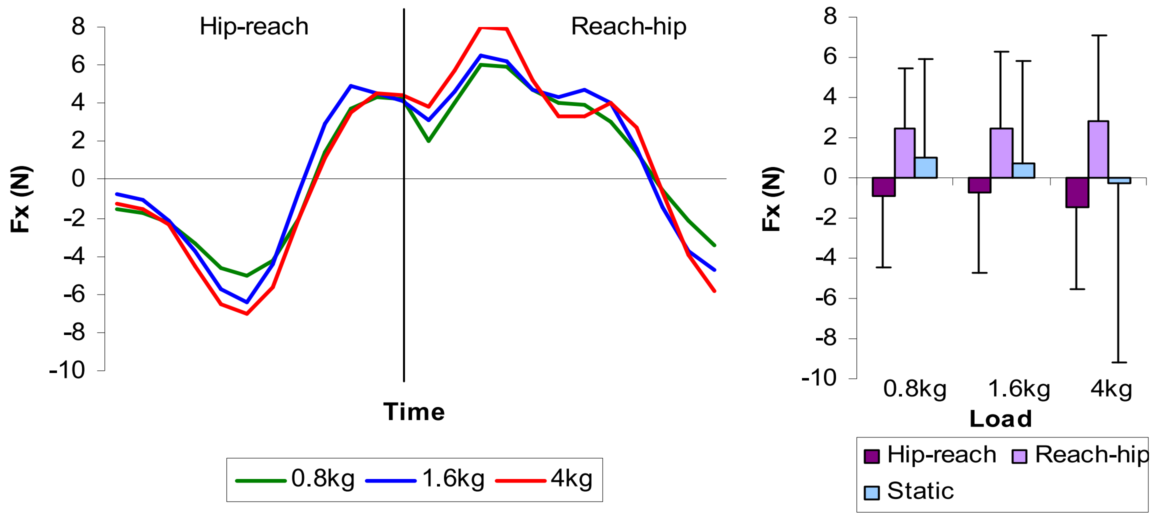
Figure 83: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the anterior/posterior direction under each loading condition during the "hip twist" task.



(A) (B)

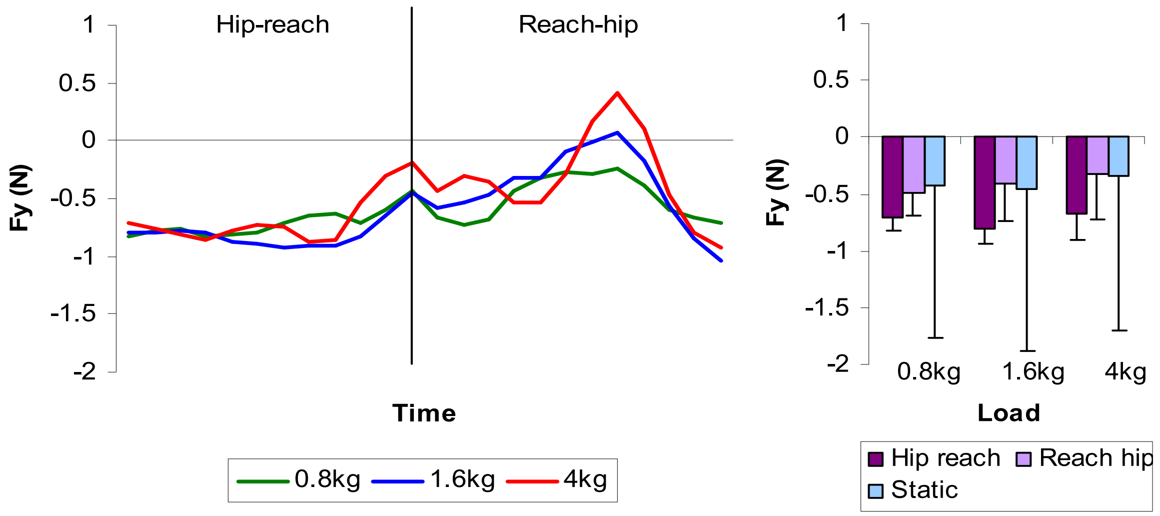
Figure 84: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the lateral direction under each loading condition during the "hip twist" task.

HIP REACH TASK



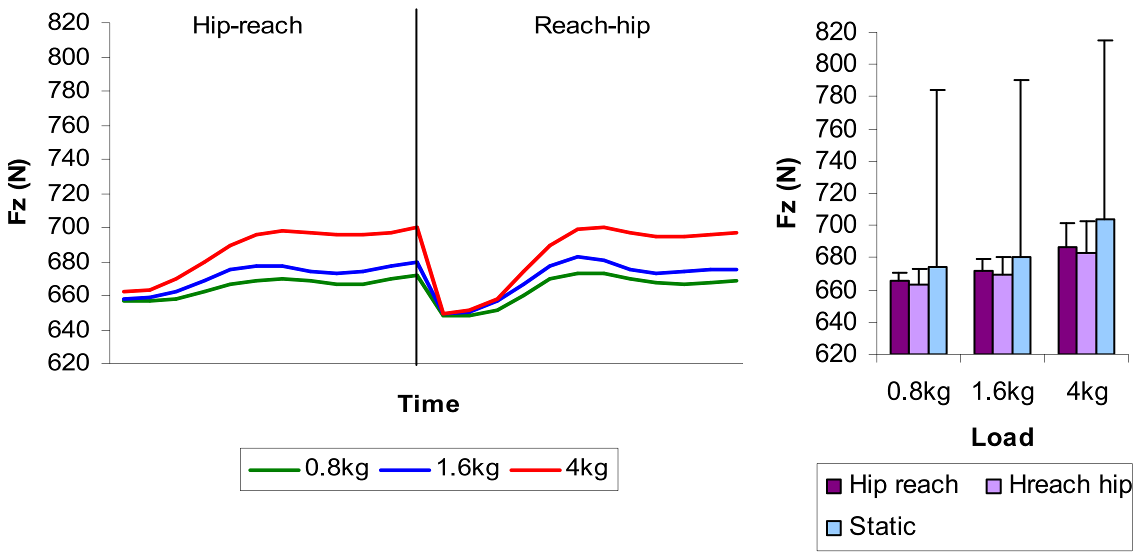
(A) (B)

Figure 85: Pattern (A) and average (B) sagittal ground reaction forces of each loading condition during the "hip reach" task.



(A) (B)

Figure 86: Pattern (A) and average (B) lateral ground reaction forces of each loading condition during the "hip reach" task.



(A) (B)

Figure 87: Pattern (A) and average (B) vertical ground reaction forces of each loading condition during the "hip reach" task.

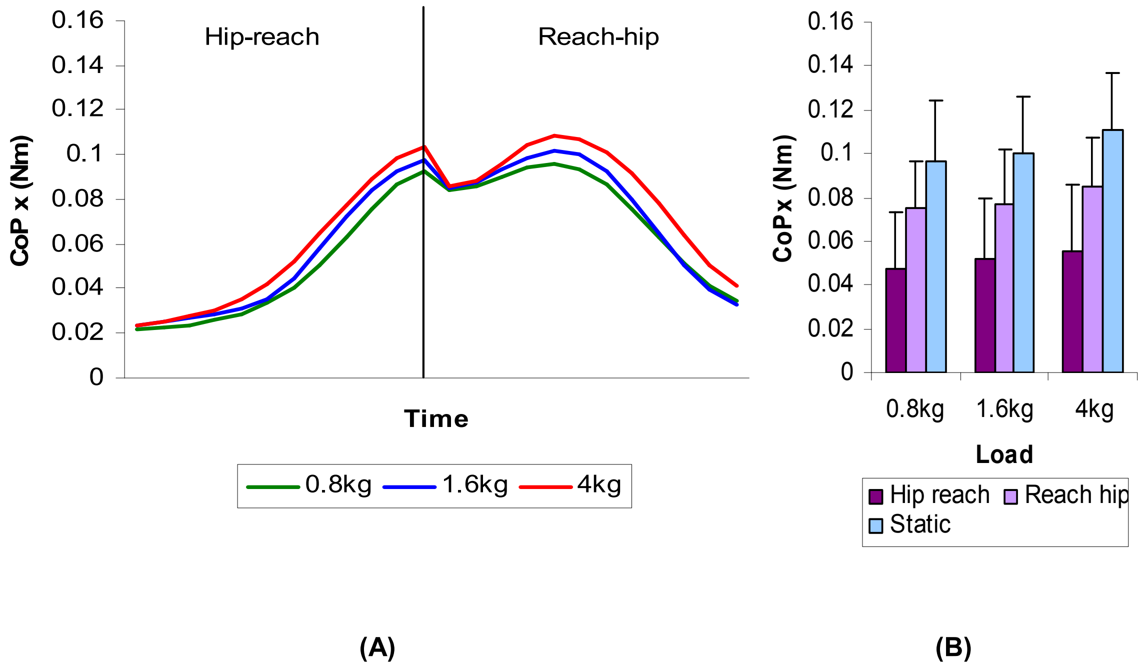


Figure 88: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the anterior/posterior direction under each loading condition in the "hip reach" task.

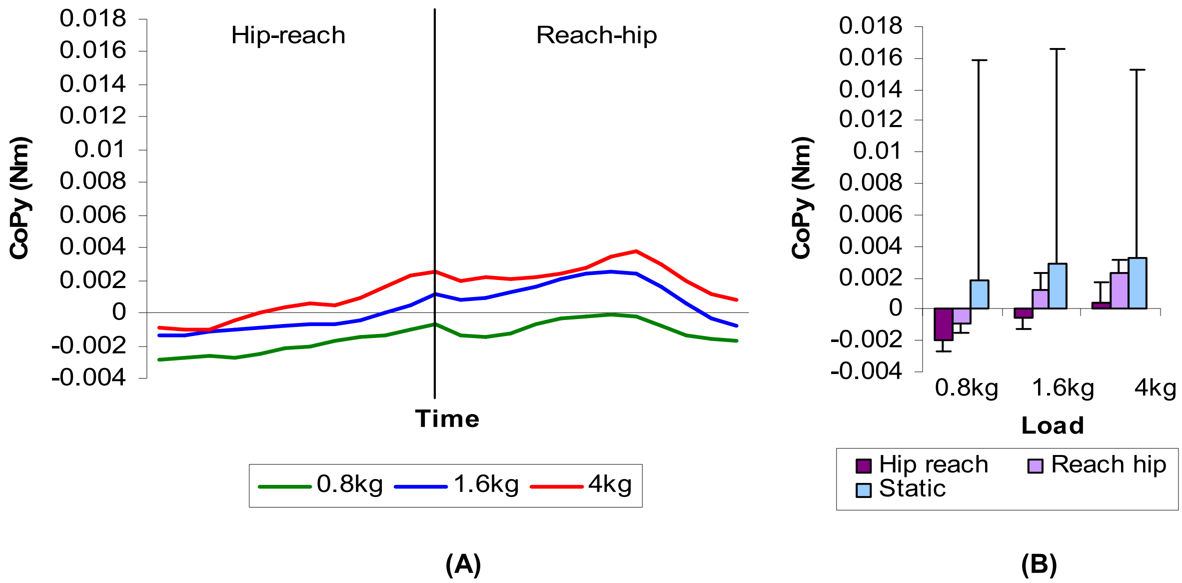


Figure 89: Pattern (A) and average (B) displacement of the centre of pressure (COP) in the lateral direction under each loading condition during the "hip reach" task.

APPENDIX D: STATISTICS TABLES

Electromyography of all tasks
- Load component
- Movement component

Ground reaction forces of all tasks
- Load component
- Movement component

Body Discomfort
- Tukey Post Hoc test

ELECTROMYOGRAPHY

LOAD COMPONENT

Hip knee task

Right lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1652.9	2	826.46	0.57804	0.563064
DATAPOI	14707.1	11	1337.01	12.15310	0.000000
CONDITIO*DATAPOI	3730.0	22	169.55	1.32754	0.142795

Left lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1287.0	2	643.48	0.43713	0.647253
DATAPOI	16204.3	11	1473.12	14.25675	0.000000
CONDITIO*DATAPOI	2618.4	22	119.02	1.06636	0.378374

Right thoracic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	3175.43	2	1587.72	2.08229	0.130609
DATAPOI	12995.68	11	1181.43	16.94051	0.000000
CONDITIO*DATAPOI	4432.18	22	201.46	2.26879	0.000755

Left thoracic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	2967.82	2	1483.91	2.74743	0.069472
DATAPOI	14614.01	11	1328.55	18.32011	0.000000
CONDITIO*DATAPOI	5128.48	22	233.11	2.91290	0.000009

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	519.234	2	259.617	3.59039	0.031608
DATAPOI	128.125	11	11.648	2.34841	0.007940
CONDITIO*DATAPOI	98.321	22	4.469	1.25899	0.189596

Left external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	189.23	2	94.613	0.58884	0.557093
DATAPOI	128.52	11	11.684	2.54780	0.003849
CONDITIO*DATAPOI	52.41	22	2.382	0.48740	0.977843

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	25.314	2	12.6570	0.71116	0.493814
DATAPOI	15.286	11	1.3896	1.95835	0.030614
CONDITIO*DATAPOI	29.899	22	1.3590	1.47623	0.072796

Right internal oblique

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	201.69	2	100.846	0.60465	0.548469
DATAPOI	686.70	11	62.427	6.92937	0.000000
CONDITIO*DATAPOI	275.91	22	12.541	1.88119	0.008350

Left internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	190.64	2	95.320	0.58607	0.558619
DATAPOI	995.96	11	90.542	7.69136	0.000000
CONDITIO*DATAPOI	336.67	22	15.303	1.32284	0.145679

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	7511.45	2	3755.723	12.50304	0.000016
DATAPOI	14668.12	11	1333.465	26.18037	0.000000
CONDITIO*DATAPOI	48642.62	22	2211.028	42.57914	0.000000

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	8028.75	2	4014.377	25.90736	0.000000
DATAPOI	11143.27	11	1013.025	34.50987	0.000000
CONDITIO*DATAPOI	34358.82	22	1561.764	52.27379	0.000000

Right lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	7012.4	2	3506.2	2.00576	0.140529
DATAPOI	6070.7	11	551.9	4.94239	0.000000
CONDITIO*DATAPOI	6063.6	22	275.6	2.26412	0.000778

Left lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4406.8	2	2203.4	1.15334	0.320200
DATAPOI	8126.2	11	738.7	9.34829	0.000000
CONDITIO*DATAPOI	4819.0	22	219.0	1.75076	0.017551

Right thoracic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	19006.5	2	9503.2	28.5747	0.000000
DATAPOI	9096.7	11	827.0	16.2744	0.000000
CONDITIO*DATAPOI	1800.3	22	81.8	1.2173	0.223183

Left thoracic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	16643.4	2	8321.7	14.9124	0.000003
DATAPOI	9488.7	11	862.6	16.4069	0.000000
CONDITIO*DATAPOI	1926.7	22	87.6	1.4076	0.100315

Right external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1621.687	2	810.844	14.36086	0.000004
DATAPOI	52.490	11	4.772	0.93710	0.504097
CONDITIO*DATAPOI	226.438	22	10.293	2.18251	0.001318

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	673.232	2	336.616	4.75411	0.010896
DATAPOI	73.952	11	6.723	2.79014	0.001558
CONDITIO*DATAPOI	58.833	22	2.674	0.96697	0.504740

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	15.516	2	7.7582	0.53525	0.587378
DATAPOI	22.655	11	2.0595	6.05389	0.000000
CONDITIO*DATAPOI	35.168	22	1.5986	2.51891	0.000142

Left rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	29.462	2	14.7308	1.25121	0.291088
DATAPOI	18.806	11	1.7096	1.58936	0.098302
CONDITIO*DATAPOI	18.349	22	0.8341	0.92933	0.555498

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	715.02	2	357.512	1.97933	0.144130
DATAPOI	298.22	11	27.111	3.52738	0.000088
CONDITIO*DATAPOI	74.80	22	3.400	0.35721	0.997399

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	995.78	2	497.89	2.77349	0.067787
DATAPOI	563.19	11	51.20	3.18986	0.000335
CONDITIO*DATAPOI	81.19	22	3.69	0.24461	0.999876

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	936.60	2	468.299	2.51380	0.086631
DATAPOI	4561.23	11	414.657	11.95739	0.000000
CONDITIO*DATAPOI	1138.53	22	51.751	1.36512	0.121319

Left latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1898.21	2	949.106	9.85415	0.000135
DATAPOI	6953.73	11	632.157	35.69666	0.000000
CONDITIO*DATAPOI	886.56	22	40.298	2.28060	0.000699

Hip reach task

Right lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	28462.1	2	14231.0	22.6949	0.000000
DATAPOI	11984.7	11	1089.5	8.0339	0.000000
CONDITIO*DATAPOI	2509.2	22	114.1	1.6879	0.024744

Left lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	22391.2	2	11195.6	19.5826	0.000000
DATAPOI	9232.1	11	839.3	7.3062	0.000000
CONDITIO*DATAPOI	2838.8	22	129.0	2.0901	0.002363

Right thorasic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	42677.2	2	21338.6	24.88047	0.000000
DATAPOI	29186.9	11	2653.4	9.42158	0.000000
CONDITIO*DATAPOI	6994.0	22	317.9	3.32294	0.000000

Left thorasic erector spiane (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	49964.3	2	24982.1	53.6903	0.000000
DATAPOI	25004.8	11	2273.2	17.5607	0.000000
CONDITIO*DATAPOI	7285.1	22	331.1	4.9839	0.000000

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4564.30	2	2282.15	15.79223	0.000001
DATAPOI	2633.83	11	239.44	21.98386	0.000000
CONDITIO*DATAPOI	2039.00	22	92.68	13.70373	0.000000

Left external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	2521.90	2	1260.952	6.34187	0.002651
DATAPOI	1755.95	11	159.632	15.76355	0.000000
CONDITIO*DATAPOI	1096.90	22	49.859	6.00203	0.000000

Right rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	785.748	2	392.874	40.66953	0.000000
DATAPOI	669.355	11	60.850	33.91663	0.000000
CONDITIO*DATAPOI	294.526	22	13.388	15.20140	0.000000

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	637.496	2	318.748	45.94330	0.000000
DATAPOI	497.922	11	45.266	32.32038	0.000000
CONDITIO*DATAPOI	188.332	22	8.561	11.33120	0.000000

Right internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4843.54	2	2421.77	27.71037	0.000000
DATAPOI	1407.65	11	127.97	11.61970	0.000000
CONDITIO*DATAPOI	840.21	22	38.19	7.19771	0.000000

Left internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	5430.77	2	2715.38	36.99195	0.000000
DATAPOI	2884.79	11	262.25	19.52378	0.000000
CONDITIO*DATAPOI	749.66	22	34.08	4.82300	0.000000

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	13408.08	2	6704.04	28.57311	0.000000
DATAPOI	21205.93	11	1927.81	32.09065	0.000000
CONDITIO*DATAPOI	2999.76	22	136.35	6.20241	0.000000

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	19466.91	2	9733.45	35.65448	0.000000
DATAPOI	24863.91	11	2260.36	24.05289	0.000000
CONDITIO*DATAPOI	4798.67	22	218.12	7.74070	0.000000

Right lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	48457.0	2	24228.5	24.8444	0.000000
DATAPOI	16737.1	11	1521.6	17.1954	0.000000
CONDITIO*DATAPOI	1597.7	22	72.6	0.9625	0.510759

Left lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	30432.2	2	15216.1	14.9419	0.000002
DATAPOI	8843.0	11	803.9	9.6035	0.000000
CONDITIO*DATAPOI	1987.9	22	90.4	1.2019	0.236651

Right thoracic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	38505.8	2	19252.9	42.4067	0.000000
DATAPOI	29918.9	11	2719.9	34.0352	0.000000
CONDITIO*DATAPOI	9756.1	22	443.5	6.4268	0.000000

Left thoracic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	30209.1	2	15104.5	25.8833	0.000000
DATAPOI	27403.1	11	2491.2	33.0325	0.000000
CONDITIO*DATAPOI	9313.8	22	423.4	6.6929	0.000000

Right external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	588.244	2	294.122	6.67199	0.001986
DATAPOI	1937.830	11	176.166	27.46724	0.000000
CONDITIO*DATAPOI	970.971	22	44.135	10.06741	0.000000

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	487.422	2	243.711	4.69274	0.011518
DATAPOI	1597.779	11	145.253	30.02623	0.000000
CONDITIO*DATAPOI	756.639	22	34.393	8.67787	0.000000

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	163.943	2	81.9714	5.18373	0.007400
DATAPOI	380.108	11	34.5553	33.91654	0.000000
CONDITIO*DATAPOI	231.509	22	10.5231	11.19227	0.000000

Left rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	231.3635	2	115.6817	16.13980	0.000001
DATAPOI	304.2310	11	27.6574	36.18625	0.000000
CONDITIO*DATAPOI	170.2036	22	7.7365	14.98487	0.000000

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	2345.84	2	1172.92	12.80921	0.000013
DATAPOI	2365.54	11	215.05	32.76346	0.000000

CONDITIO*DATAPOI	387.13	22	17.60	2.59834	0.000082

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	2598.09	2	1299.04	15.27749	0.000002
DATAPOI	2199.87	11	199.99	19.52820	0.000000
CONDITIO*DATAPOI	652.05	22	29.64	4.44423	0.000000

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4155.76	2	2077.88	40.64719	0.000000
DATAPOI	12770.09	11	1160.92	42.58248	0.000000
CONDITIO*DATAPOI	3127.02	22	142.14	19.26473	0.000000

Left latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4291.07	2	2145.54	18.55281	0.000000
DATAPOI	15667.99	11	1424.36	58.58448	0.000000
CONDITIO*DATAPOI	4358.40	22	198.11	16.49744	0.000000

Hip shoulder task

Right lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	23288.0	2	11644.0	58.2476	0.000000
DATAPOI	8630.4	11	784.6	32.1934	0.000000
CONDITIO*DATAPOI	1875.1	22	85.2	4.1904	0.000000

Left lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	29863.0	2	14931.5	39.1732	0.000000
DATAPOI	10370.0	11	942.7	39.4772	0.000000
CONDITIO*DATAPOI	2007.9	22	91.3	4.6955	0.000000

Right thorasic erector spiane (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	40856.9	2	20428.5	61.9260	0.000000
DATAPOI	15029.7	11	1366.3	48.4705	0.000000
CONDITIO*DATAPOI	4664.9	22	212.0	8.7734	0.000000

Left thorasic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	37590.7	2	18795.3	83.8373	0.000000
DATAPOI	10923.1	11	993.0	37.4647	0.000000
CONDITIO*DATAPOI	3186.8	22	144.9	5.7260	0.000000

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1541.489	2	770.745	9.92386	0.000127
DATAPOI	471.660	11	42.878	18.99874	0.000000
CONDITIO*DATAPOI	125.447	22	5.702	2.05840	0.002876

Left external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	527.591	2	263.795	8.40661	0.000450
DATAPOI	353.651	11	32.150	17.14345	0.000000
CONDITIO*DATAPOI	183.526	22	8.342	4.32028	0.000000

Right rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	70.43	2	35.215	0.23475	0.791248
DATAPOI	117.85	11	10.714	2.52854	0.004132
CONDITIO*DATAPOI	898.37	22	40.835	0.91871	0.569938

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	265.241	2	132.620	11.72839	0.000030
DATAPOI	117.130	11	10.648	6.51840	0.000000
CONDITIO*DATAPOI	69.888	22	3.177	1.39190	0.107683

Right internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1984.219	2	992.110	60.8734	0.000000
DATAPOI	1365.520	11	124.138	76.6627	0.000000
CONDITIO*DATAPOI	311.752	22	14.171	10.1711	0.000000

Left internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	2024.66	2	1012.33	26.3929	0.000000
DATAPOI	1657.68	11	150.70	52.5384	0.000000
CONDITIO*DATAPOI	297.28	22	13.51	6.8339	0.000000

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	6165.64	2	3082.82	35.29597	0.000000
DATAPOI	6569.23	11	597.20	35.61001	0.000000
CONDITIO*DATAPOI	1690.52	22	76.84	7.60902	0.000000

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	8660.79	2	4330.40	35.1105	0.000000
DATAPOI	9813.00	11	892.09	37.9151	0.000000
CONDITIO*DATAPOI	3726.05	22	169.37	13.2408	0.000000

Right lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	11690.47	2	5845.23	26.2282	0.000000
DATAPOI	9020.78	11	820.07	55.7229	0.000000
CONDITIO*DATAPOI	1958.06	22	89.00	6.1094	0.000000

Left lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	15395.43	2	7697.72	23.7168	0.000000
DATAPOI	9661.07	11	878.28	57.9938	0.000000
CONDITIO*DATAPOI	2019.69	22	91.80	6.4011	0.000000

Right thorasic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	14735.9	2	7368.0	29.9146	0.000000
DATAPOI	18910.1	11	1719.1	92.8834	0.000000
CONDITIO*DATAPOI	4485.4	22	203.9	12.6242	0.000000

Left thorasic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	14537.1	2	7268.6	30.3422	0.000000
DATAPOI	17116.6	11	1556.1	90.1244	0.000000
CONDITIO*DATAPOI	2424.8	22	110.2	7.5561	0.000000

Right external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	701.473	2	350.737	6.38915	0.002543
DATAPOI	231.478	11	21.043	12.22951	0.000000
CONDITIO*DATAPOI	79.657	22	3.621	1.86492	0.009180

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	178.180	2	89.090	2.87172	0.061803
DATAPOI	233.496	11	21.227	15.91925	0.000000
CONDITIO*DATAPOI	66.912	22	3.041	2.25779	0.000811

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	138.783	2	69.3916	5.72346	0.004573
DATAPOI	83.211	11	7.5646	6.89409	0.000000
CONDITIO*DATAPOI	79.368	22	3.6076	1.17976	0.256955

Left rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	66.577	2	33.2884	4.64541	0.012023
DATAPOI	86.017	11	7.8197	8.92886	0.000000
CONDITIO*DATAPOI	48.155	22	2.1888	1.51595	0.060051

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	808.541	2	404.271	28.67648	0.000000
DATAPOI	1030.855	11	93.714	68.07328	0.000000
CONDITIO*DATAPOI	253.846	22	11.538	14.94465	0.000000

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	898.934	2	449.467	15.07336	0.000002
DATAPOI	1321.502	11	120.137	55.85124	0.000000
CONDITIO*DATAPOI	333.263	22	15.148	11.73262	0.000000

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1246.44	2	623.22	23.37592	0.000000
DATAPOI	6484.88	11	589.53	46.63183	0.000000
CONDITIO*DATAPOI	1072.52	22	48.75	15.42556	0.000000

Left Latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	1657.54	2	828.77	15.88164	0.000001
DATAPOI	8034.83	11	730.44	75.32090	0.000000
CONDITIO*DATAPOI	1616.11	22	73.46	15.25138	0.000000

Hip twist task

Right lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	944.54	2	472.272	1.87133	0.159859
DATAPOI	3933.63	11	357.603	38.12477	0.000000
CONDITIO*DATAPOI	2106.64	22	95.756	9.91235	0.000000

Left lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	7185.2	2	3592.6	17.3731	0.000000
DATAPOI	750.1	11	68.2	3.8691	0.000022
CONDITIO*DATAPOI	861.2	22	39.1	1.8839	0.008218

Right thoracic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	3471.21	2	1735.60	7.28962	0.001164
DATAPOI	4479.47	11	407.22	26.67406	0.000000
CONDITIO*DATAPOI	1461.20	22	66.42	4.93057	0.000000

Left thoracic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p

CONDITIO	13763.25	2	6881.63	19.1492	0.000000
DATAPOI	317.46	11	28.86	2.6153	0.003000
CONDITIO*DATAPOI	444.04	22	20.18	1.2992	0.160930

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	460.36	2	230.180	1.20914	0.303254
DATAPOI	2204.48	11	200.407	8.36554	0.000000
CONDITIO*DATAPOI	167.06	22	7.594	0.58016	0.938334

Left external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	5904.24	2	2952.12	6.48429	0.002340
DATAPOI	18751.81	11	1704.71	37.14655	0.000000
CONDITIO*DATAPOI	3079.29	22	139.97	6.07003	0.000000

Right rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	12.8570	2	6.4285	0.99849	0.372479
DATAPOI	26.7957	11	2.4360	4.30885	0.000004
CONDITIO*DATAPOI	9.0445	22	0.4111	0.65193	0.887330

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	30.1592	2	15.0796	1.64217	0.199307
DATAPOI	108.6974	11	9.8816	9.72714	0.000000
CONDITIO*DATAPOI	12.8586	22	0.5845	0.66279	0.877995

Right internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	96.530	2	48.265	1.18277	0.311147
DATAPOI	80.147	11	7.286	3.24249	0.000272
CONDITIO*DATAPOI	63.649	22	2.893	1.87841	0.008487

Left internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	3059.08	2	1529.54	16.8535	0.000001
DATAPOI	2418.18	11	219.83	29.6190	0.000000
CONDITIO*DATAPOI	370.87	22	16.86	3.6666	0.000000

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	446.096	2	223.048	3.82717	0.025398
DATAPOI	42.876	11	3.898	0.79144	0.648789
CONDITIO*DATAPOI	83.477	22	3.794	1.47707	0.072504

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p

CONDITIO	1541.25	2	770.63	21.37573	0.000000
DATAPOI	2496.36	11	226.94	33.68607	0.000000
CONDITIO*DATAPOI	262.01	22	11.91	4.14997	0.000000

Right lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4208.81	2	2104.41	9.93696	0.000126
DATAPOI	4019.04	11	365.37	39.44967	0.000000
CONDITIO*DATAPOI	2280.07	22	103.64	11.88302	0.000000

Left lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	18146.71	2	9073.35	41.4107	0.000000
DATAPOI	1423.07	11	129.37	9.4698	0.000000
CONDITIO*DATAPOI	525.36	22	23.88	1.7163	0.021213

Right thorasic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4191.28	2	2095.64	11.90112	0.000026
DATAPOI	5993.42	11	544.86	36.31203	0.000000
CONDITIO*DATAPOI	2218.35	22	100.83	7.91627	0.000000

Left thorasic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	20190.56	2	10095.28	51.3982	0.000000
DATAPOI	903.99	11	82.18	5.5807	0.000000
CONDITIO*DATAPOI	490.29	22	22.29	2.1295	0.001844

Right external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	99.095	2	49.5475	0.83241	0.438321
DATAPOI	173.219	11	15.7472	2.93252	0.000906
CONDITIO*DATAPOI	217.024	22	9.8647	2.79150	0.000021

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	3218.13	2	1609.06	8.68422	0.000356
DATAPOI	15431.39	11	1402.85	34.13167	0.000000
CONDITIO*DATAPOI	4417.90	22	200.81	8.21910	0.000000

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	16.6431	2	8.32154	3.68000	0.029093
DATAPOI	8.8385	11	0.80350	4.24073	0.000005
CONDITIO*DATAPOI	3.8581	22	0.17537	0.76187	0.774938

Left rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	36.6465	2	18.3233	5.84141	0.004119
DATAPOI	46.3241	11	4.2113	9.14384	0.000000
CONDITIO*DATAPOI	26.4071	22	1.2003	2.64867	0.000058

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	159.700	2	79.8502	3.72999	0.027780
DATAPOI	44.386	11	4.0351	3.71080	0.000042
CONDITIO*DATAPOI	18.486	22	0.8403	0.92770	0.557707

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	3592.88	2	1796.44	27.71511	0.000000
DATAPOI	2963.21	11	269.38	33.39531	0.000000
CONDITIO*DATAPOI	632.91	22	28.77	6.08544	0.000000

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	225.055	2	112.5275	7.68750	0.000827
DATAPOI	90.090	11	8.1900	5.32299	0.000000
CONDITIO*DATAPOI	82.244	22	3.7383	2.11917	0.001969

Left latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	860.348	2	430.174	43.8906	0.000000
DATAPOI	288.750	11	26.250	14.5332	0.000000
CONDITIO*DATAPOI	127.357	22	5.789	5.4496	0.000000

MOVEMENT COMPONENT

Hip knee task

Right lumbar erector spiane (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	230.69	2	115.347	0.403671	0.669067
DATAPOI	6860.24	11	623.658	8.233420	0.000000
LOADS*DATAPOI	1377.91	22	62.632	8.248441	0.000000

Left lumbar erector spinae

	SS	Degr. of - Freedom	MS	F	p
LOADS	3.06	2	1.529	0.007048	0.992977
DATAPOI	7585.18	11	689.562	8.925227	0.000000
LOADS*DATAPOI	1417.35	22	64.425	9.738352	0.000000

Right thorasic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	96.88	2	48.442	0.39396	0.675537
DATAPOI	3959.73	11	359.976	7.29582	0.000000
LOADS*DATAPOI	1259.22	22	57.237	10.59992	0.000000

Left thorasic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	918.99	2	459.495	2.254987	0.110769
DATAPOI	4671.24	11	424.658	8.692760	0.000000
LOADS*DATAPOI	1926.27	22	87.558	9.450116	0.000000

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	32.437	2	16.219	1.86552	0.160752
DATAPOI	124.234	11	11.294	6.16244	0.000000
LOADS*DATAPOI	12.474	22	0.567	1.96107	0.005204

Left external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	49.068	2	24.534	4.36147	0.015563
DATAPOI	312.878	11	28.443	7.18318	0.000000
LOADS*DATAPOI	10.681	22	0.485	1.63183	0.033344

Right rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1.1295	2	0.56473	1.20489	0.304514
DATAPOI	6.7701	11	0.61547	4.19173	0.000006
LOADS*DATAPOI	1.4181	22	0.06446	1.70334	0.022767

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	2.8632	2	1.4316	1.73717	0.181867
DATAPOI	9.3512	11	0.8501	5.84103	0.000000
LOADS*DATAPOI	2.2776	22	0.1035	2.10652	0.002132

Right internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	76.522	2	38.261	3.73736	0.027591
DATAPOI	1099.203	11	99.928	19.04330	0.000000
LOADS*DATAPOI	67.422	22	3.065	5.81535	0.000000

Left internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	10.288	2	5.144	0.37378	0.689194
DATAPOI	1071.598	11	97.418	18.06314	0.000000
LOADS*DATAPOI	93.987	22	4.272	5.69741	0.000000

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	6.635	2	3.318	0.16623	0.847114
DATAPOI	1402.237	11	127.476	28.36868	0.000000
LOADS*DATAPOI	177.274	22	8.058	9.11823	0.000000

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	24.294	2	12.147	1.34460	0.265828
DATAPOI	1179.485	11	107.226	35.09527	0.000000
LOADS*DATAPOI	149.562	22	6.798	11.95191	0.000000

Right lower erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	3997.8	2	1998.9	4.5195	0.013478
DATAPOI	35620.8	11	3238.3	37.7388	0.000000
LOADS*DATAPOI	668.9	22	30.4	4.1008	0.000000

Lower erector spinae left (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	113272.3	2	56636.2	139.4904	0.000000
DATAPOI	16180.7	11	1471.0	32.3586	0.000000
LOADS*DATAPOI	7751.8	22	352.4	13.9514	0.000000

Right thorasic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	2814.6	2	1407.3	8.53178	0.000405
DATAPOI	3229.9	11	293.6	7.88626	0.000000
LOADS*DATAPOI	746.3	22	33.9	9.00234	0.000000

Left thoracic erector spiane (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	3647.1	2	1823.5	8.27035	0.000504
DATAPOI	3406.1	11	309.6	7.59555	0.000000
LOADS*DATAPOI	812.5	22	36.9	9.91463	0.000000

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	77.095	2	38.548	5.09346	0.008024
DATAPOI	69.023	11	6.275	2.96647	0.000796
LOADS*DATAPOI	12.443	22	0.566	1.65865	0.028943

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	108.720	2	54.360	6.36173	0.002605
DATAPOI	106.854	11	9.714	8.70033	0.000000
LOADS*DATAPOI	8.191	22	0.372	2.18142	0.001327

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	9.4388	2	4.7194	8.96277	0.000282
DATAPOI	16.9881	11	1.5444	19.44958	0.000000
LOADS*DATAPOI	3.3588	22	0.1527	5.32822	0.000000

Left rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	17.2163	2	8.6081	12.40416	0.000017
DATAPOI	17.3870	11	1.5806	8.64225	0.000000
LOADS*DATAPOI	2.2416	22	0.1019	1.36197	0.123013

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	259.99	2	130.00	11.44542	0.000037
DATAPOI	711.54	11	64.69	11.94657	0.000000
LOADS*DATAPOI	26.02	22	1.18	2.17576	0.001376

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	384.10	2	192.05	12.88438	0.000012
DATAPOI	1066.20	11	96.93	10.63264	0.000000
LOADS*DATAPOI	44.38	22	2.02	2.10451	0.002159

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	253.498	2	126.749	6.02049	0.003516
DATAPOI	3130.101	11	284.555	26.68823	0.000000
LOADS*DATAPOI	410.369	22	18.653	8.20815	0.000000

Left latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	302.075	2	151.038	12.0798	0.000023
DATAPOI	1960.509	11	178.228	24.6298	0.000000
LOADS*DATAPOI	592.728	22	26.942	24.5689	0.000000

Hip shoulder task

Right lumbar erector spinae (condition **1)**

	SS	Degr. of - Freedom	MS	F	p
LOADS	1671.49	2	835.744	23.0900	0.000000
DATAPOI	20777.20	11	1888.836	248.6364	0.000000
LOADS*DATAPOI	776.93	22	35.315	24.6663	0.000000

Left lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	753.09	2	376.546	15.5380	0.000002
DATAPOI	22302.00	11	2027.454	315.8933	0.000000
LOADS*DATAPOI	924.28	22	42.013	28.1372	0.000000

Right thoracic erector spinae (condition1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	576.79	2	288.397	8.6746	0.000359
DATAPOI	24109.30	11	2191.755	251.3872	0.000000
LOADS*DATAPOI	1456.13	22	66.188	36.5065	0.000000

Left thoracic erector spinae (condition1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	301.70	2	150.85	7.9113	0.000684
DATAPOI	25480.08	11	2316.37	223.6184	0.000000
LOADS*DATAPOI	1027.87	22	46.72	26.5599	0.000000

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	20.572	2	10.2860	0.65545	0.521669
DATAPOI	40.466	11	3.6788	6.10711	0.000000
LOADS*DATAPOI	44.179	22	2.0081	12.05322	0.000000

Left external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	10.3815	2	5.1907	0.66341	0.517594
DATAPOI	29.2686	11	2.6608	6.17492	0.000000
LOADS*DATAPOI	37.8230	22	1.7192	13.55652	0.000000

Right rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOAD	24.667	2	12.33357	2.818378	0.064982
DATAPOI	54.576	11	4.96148	1.338889	0.199369
LOAD*DATAPOI	35.012	22	1.59145	1.229939	0.212598

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	9.0942	2	4.547113	3.565847	0.032335
DATAPOI	16.8674	11	1.533400	7.798506	0.000000
LOADS*DATAPOI	12.9819	22	0.590086	4.414647	0.000000

Right internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	50.3496	2	25.17481	10.1872	0.000103
DATAPOI	776.0263	11	70.54785	138.0639	0.000000
LOADS*DATAPOI	126.3932	22	5.74515	53.7655	0.000000

Left internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	48.330	2	24.1651	5.7745	0.004370
DATAPOI	1122.378	11	102.0344	134.6008	0.000000
LOADS*DATAPOI	149.016	22	6.7735	39.5100	0.000000

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	138.229	2	69.115	2.28317	0.107836
DATAPOI	4547.295	11	413.390	60.45173	0.000000
LOADS*DATAPOI	622.352	22	28.289	27.48769	0.000000

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	988.289	2	494.144	38.53560	0.000000
DATAPOI	2427.590	11	220.690	55.96105	0.000000
LOADS*DATAPOI	5341.222	22	242.783	77.59720	0.000000

Right lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	3861.71	2	1930.85	43.0497	0.000000
DATAPOI	7665.11	11	696.83	140.6770	0.000000
LOADS*DATAPOI	832.72	22	37.85	38.8039	0.000000

Left lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	2756.66	2	1378.33	43.3766	0.000000
DATAPOI	8482.88	11	771.17	135.1584	0.000000
LOADS*DATAPOI	887.10	22	40.32	39.5487	0.000000

Right thoracic erector spinae (condition2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	3861.32	2	1930.66	48.0799	0.000000
DATAPOI	12365.24	11	1124.11	153.3401	0.000000
LOADS*DATAPOI	1772.57	22	80.57	69.3187	0.000000

Left thoracic erector spiane (condition2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	2909.05	2	1454.52	45.4177	0.000000
DATAPOI	12624.31	11	1147.66	172.5633	0.000000
LOADS*DATAPOI	1495.88	22	67.99	64.5114	0.000000

Right external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	75.272	2	37.63576	2.53135	0.085204
DATAPOI	79.775	11	7.25230	22.92646	0.000000
LOADS*DATAPOI	22.690	22	1.03137	8.18682	0.000000

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	49.8554	2	24.92769	3.44139	0.036292
DATAPOI	86.6525	11	7.87750	24.90310	0.000000
LOADS*DATAPOI	22.1623	22	1.00738	10.92516	0.000000

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	33.5489	2	16.77446	12.05377	0.000023
DATAPOI	34.6106	11	3.14641	5.81899	0.000000
LOADS*DATAPOI	9.5362	22	0.43346	2.98483	0.000005

Left rectus abdominis (conditon2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	43.2556	2	21.62781	14.66269	0.000003
DATAPOI	28.1722	11	2.56110	22.23446	0.000000
LOADS*DATAPOI	8.9769	22	0.40804	5.25433	0.000000

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	248.596	2	124.298	52.18625	0.000000
DATAPOI	313.743	11	28.522	66.49545	0.000000
LOADS*DATAPOI	97.066	22	4.412	56.71279	0.000000

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	279.144	2	139.572	24.28652	0.000000
DATAPOI	414.712	11	37.701	64.61981	0.000000
LOADS*DATAPOI	125.018	22	5.683	44.61122	0.000000

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1504.23	2	752.12	17.92043	0.000000
DATAPOI	2444.08	11	222.19	55.23595	0.000000
LOADS*DATAPOI	578.06	22	26.28	43.58721	0.000000

Left latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	13966.75	2	6983.376	89.08924	0.000000
DATAPOI	365.39	11	33.217	17.44685	0.000000
LOADS*DATAPOI	8898.87	22	404.494	84.98530	0.000000

Hip twist task

Right lumbar erector spiane (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	132.89	2	66.44	1.97447	0.144803
DATAPOI	881.60	11	80.15	14.75514	0.000000
LOADS*DATAPOI	429.41	22	19.52	31.37299	0.000000

Left lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	457.19	2	228.593	3.5595	0.032526
DATAPOI	13602.52	11	1236.593	100.3574	0.000000
LOADS*DATAPOI	95.32	22	4.333	3.5139	0.000000

Right thoracic erector spinae (condition1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	698.74	2	349.37	9.05579	0.000261
DATAPOI	1243.44	11	113.04	17.90129	0.000000
LOADS*DATAPOI	439.79	22	19.99	20.81700	0.000000

Left thoracic erector spiane (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	71.700	2	35.8501	1.40638	0.250361
DATAPOI	1506.117	11	136.9197	24.32719	0.000000
LOADS*DATAPOI	42.289	22	1.9222	2.11663	0.002001

Right external oblique (conditon1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	50.92	2	25.4591	1.38171	0.256424
DATAPOI	9972.22	11	906.5658	35.84304	0.000000
LOADS*DATAPOI	172.75	22	7.8521	5.99773	0.000000

Left external oblique (condiiton1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	6556.82	2	3278.41	26.39099	0.000000
DATAPOI	23225.95	11	2111.45	81.63392	0.000000
LOADS*DATAPOI	1850.33	22	84.11	28.23632	0.000000

Right rectus abdominis (conditon1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.0601	2	0.03006	0.05046	0.950819
DATAPOI	190.3558	11	17.30508	48.34893	0.000000
LOADS*DATAPOI	2.1222	22	0.09646	2.76209	0.000026

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	29.1913	2	14.59564	8.90077	0.000297
DATAPOI	266.1425	11	24.19478	36.01994	0.000000
LOADS*DATAPOI	8.0971	22	0.36805	6.18297	0.000000

Right internal oblique (condition1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	30.8323	2	15.4162	3.64294	0.030107
DATAPOI	604.8138	11	54.9831	69.91497	0.000000
LOADS*DATAPOI	8.1780	22	0.3717	2.83453	0.000015

Left internal oblique (condition1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	852.337	2	426.168	15.98438	0.000001
DATAPOI	3287.709	11	298.883	81.17314	0.000000
LOADS*DATAPOI	209.332	22	9.515	22.43527	0.000000

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	13.523	2	6.7613	1.04226	0.356874
CONDITIO	563.205	11	51.2005	21.81488	0.000000
LOADS*CONDITIO	5.560	22	0.2527	1.01067	0.447411

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	18.335	2	9.167	0.9037	0.408693
DATAPOI	1941.786	11	176.526	116.5507	0.000000
LOADS*DATAPOI	210.983	22	9.590	25.7029	0.000000

Right lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	609.14	2	304.57	8.26830	0.000505
DATAPOI	1994.43	11	181.31	38.14788	0.000000
LOADS*DATAPOI	441.23	22	20.06	33.68464	0.000000

Left lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	608.95	2	304.47	3.85727	0.024703
DATAPOI	6914.42	11	628.58	84.47669	0.000000
LOADS*DATAPOI	130.27	22	5.92	6.54081	0.000000

Right thoracic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1343.29	2	671.647	14.97112	0.000002
DATAPOI	2888.43	11	262.585	54.38773	0.000000
LOADS*DATAPOI	606.04	22	27.547	30.61602	0.000000

Let thoracic erector spiane (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	179.589	2	89.795	2.98297	0.055672
DATAPOI	2960.141	11	269.104	57.36441	0.000000
LOADS*DATAPOI	90.993	22	4.136	4.78201	0.000000

Right external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	47.034	2	23.517	1.47366	0.234560
DATAPOI	458.686	11	41.699	34.55049	0.000000
LOADS*DATAPOI	22.917	22	1.042	3.14542	0.000002

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	14075.3	2	7037.6	44.43362	0.000000
DATAPOI	7702.6	11	700.2	46.67290	0.000000
LOADS*DATAPOI	1489.1	22	67.7	26.43499	0.000000

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1.7043	2	0.8521	1.28583	0.281449
DATAPOI	11.5116	11	1.0465	24.72346	0.000000
LOADS*DATAPOI	0.8674	22	0.0394	2.91957	0.000008

Left rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	51.8695	2	25.9347	14.44661	0.000004
DATAPOI	32.8476	11	2.9861	26.85718	0.000000
LOADS*DATAPOI	5.0252	22	0.2284	7.89424	0.000000

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	15.5035	2	7.75177	2.01976	0.138657
DATAPOI	68.0520	11	6.18655	10.90148	0.000000
LOADS*DATAPOI	4.2170	22	0.19168	2.71548	0.000036

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1737.52	2	868.76	28.70473	0.000000
DATAPOI	2344.36	11	213.12	65.65804	0.000000
LOADS*DATAPOI	273.40	22	12.43	26.78020	0.000000

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	3.810	2	1.9049	0.46841	0.627513
DATAPOI	4.252	11	0.3865	0.90292	0.537337
LOADS*DATAPOI	10.041	22	0.4564	4.36730	0.000000

Left latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	281.741	2	140.871	23.1390	0.000000
DATAPOI	273.720	11	24.884	69.7177	0.000000
LOADS*DATAPOI	29.809	22	1.355	13.1378	0.000000

Hip reach task

Right lumbar erector spiane (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	6586.50	2	3293.25	29.2415	0.000000
DATAPOI	49986.89	11	4544.26	113.2285	0.000000
LOADS*DATAPOI	1069.29	22	48.60	7.3263	0.000000

Left lumbar erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	7236.39	2	3618.19	26.3210	0.000000
DATAPOI	44829.72	11	4075.43	101.0857	0.000000
LOADS*DATAPOI	889.00	22	40.41	7.2025	0.000000

Right thoracic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	6602.16	2	3301.08	8.3399	0.000476
DATAPOI	53342.12	11	4849.28	123.3979	0.000000
LOADS*DATAPOI	2660.58	22	120.94	10.3958	0.000000

Left thoracic erector spinae (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	7556.70	2	3778.35	19.1138	0.000000
DATAPOI	52954.89	11	4814.08	109.2617	0.000000
LOADS*DATAPOI	2387.31	22	108.51	17.4789	0.000000

Right external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1268.309	2	634.1546	19.24503	0.000000
DATAPOI	1220.147	11	110.9224	40.14657	0.000000
LOADS*DATAPOI	316.999	22	14.4091	21.31486	0.000000

Left external oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1204.017	2	602.0086	13.92583	0.000005
DATAPOI	1050.222	11	95.4747	57.07940	0.000000
LOADS*DATAPOI	198.975	22	9.0443	14.52494	0.000000

Right rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	50.6677	2	25.33384	14.80379	0.000003
DATAPOI	340.4852	11	30.95320	61.31525	0.000000
LOADS*DATAPOI	70.1291	22	3.18769	30.30346	0.000000

Left rectus abdominis (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	42.0723	2	21.03615	9.10152	0.000251
DATAPOI	301.1003	11	27.37275	68.20092	0.000000
LOADS*DATAPOI	50.4878	22	2.29490	29.62737	0.000000

Right internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	621.981	2	310.990	14.71221	0.000003
DATAPOI	4068.960	11	369.905	78.83535	0.000000
LOADS*DATAPOI	155.555	22	7.071	13.07963	0.000000

Left internal oblique (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	597.949	2	298.975	11.06936	0.000050
DATAPOI	4325.244	11	393.204	61.30335	0.000000
LOADS*DATAPOI	264.774	22	12.035	18.12249	0.000000

Right latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1225.36	2	612.678	9.86110	0.000134
DATAPOI	23361.16	11	2123.742	59.30962	0.000000
LOADS*DATAPOI	1848.49	22	84.022	29.10595	0.000000

Left latissimus dorsi (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1172.27	2	586.14	8.07280	0.000596
DATAPOI	31301.63	11	2845.60	79.10128	0.000000
LOADS*DATAPOI	2262.08	22	102.82	21.98465	0.000000

Right lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	4965.1	2	2482.55	12.5910	0.000015
DATAPOI	94547.4	11	8595.22	159.4242	0.000000
LOADS*DATAPOI	1372.1	22	62.37	11.0013	0.000000

Left lumbar erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	5822.1	2	2911.07	14.3303	0.000004
DATAPOI	86810.5	11	7891.87	163.7535	0.000000
LOADS*DATAPOI	776.8	22	35.31	6.2160	0.000000

Right thoracic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	12496.0	2	6248.0	15.6493	0.000001
DATAPOI	77266.2	11	7024.2	156.0437	0.000000
LOADS*DATAPOI	2942.8	22	133.8	25.0354	0.000000

Left thoracic erector spinae (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	14659.74	2	7329.87	30.6368	0.000000
DATAPOI	76092.57	11	6917.51	142.2579	0.000000
LOADS*DATAPOI	2712.56	22	123.30	25.1010	0.000000

Right external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	2673.440	2	1336.720	37.32014	0.000000
DATAPOI	311.908	11	28.355	22.00229	0.000000
LOADS*DATAPOI	208.192	22	9.463	23.81726	0.000000

Left external oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	2101.445	2	1050.723	28.03652	0.000000
DATAPOI	246.331	11	22.394	26.06215	0.000000
LOADS*DATAPOI	168.974	22	7.681	23.77405	0.000000

Right rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	161.4302	2	80.7151	47.01620	0.000000
DATAPOI	159.5266	11	14.5024	59.95159	0.000000
LOADS*DATAPOI	42.4193	22	1.9282	26.96004	0.000000

Left rectus abdominis (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	125.2027	2	62.6014	31.39454	0.000000
DATPOIN	141.6456	11	12.8769	74.21586	0.000000
LOADS*DATPOIN	33.1406	22	1.5064	32.51302	0.000000

Right internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	867.654	2	433.827	15.15225	0.000002
DATAPOI	3967.911	11	360.719	80.51889	0.000000
LOADS*DATAPOI	208.346	22	9.470	22.43947	0.000000

Left internal oblique (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	899.228	2	449.614	13.62174	0.000007
DATAPOI	4712.818	11	428.438	49.43739	0.000000
LOADS*DATAPOI	209.405	22	9.518	16.07057	0.000000

Right latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	5788.38	2	2894.19	45.21601	0.000000
DATAPOI	8360.08	11	760.01	72.93363	0.000000
LOADS*DATAPOI	1203.91	22	54.72	40.83285	0.000000

Left latissimus dorsi (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	6304.60	2	3152.30	60.52355	0.000000
DATAPOI	11541.07	11	1049.19	76.83906	0.000000
LOADS*DATAPOI	1508.92	22	68.59	48.86587	0.000000

GROUND REACTION FORCES

LOAD COMPONENT

Hip knee task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITOI	290.2	2	145.1156	1.977717	0.144353
DATAPOI	7098.7	11	645.3385	3.348141	0.000180
CONDITOI*DATAPOI	10171.4	22	462.3344	2.833330	0.000016

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.04	2	0.02139	0.000760	0.999240
DATAPOI	459.46	11	41.76890	2.627399	0.002868
CONDITIO*DATAPOI	219.45	22	9.97504	0.783148	0.749247

Fz (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITOI	298812	2	149406	36.2222	0.000000
DATAPOI	294965	11	26815	19.1622	0.000000
CONDITOI*DATAPOI	103828	22	4719	3.7852	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITOI	0.066511	2	0.033256	2.49872	0.087878
DATAPOI	0.024325	11	0.002211	1.83128	0.046402
CONDITOI*DATAPOI	0.100757	22	0.004580	4.02898	0.000000

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.003841	2	0.001920	0.299058	0.742250
DATAPOI	0.006449	11	0.000586	5.150999	0.000000
CONDITIO*DATAPOI	0.003048	22	0.000139	1.115130	0.322701

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	581.8	2	290.8918	3.404720	0.037551
DATAPOI	9525.8	11	865.9837	8.279653	0.000000
CONDITIO*DATAPOI	12325.1	22	560.2307	4.002812	0.000000

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	2.43	2	1.21289	0.051150	0.950163
DATAPOI	121.58	11	11.05270	0.740536	0.699555
CONDITIO*DATAPOI	416.50	22	18.93175	1.214366	0.225734

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	399273	2	199637	50.8347	0.000000
DATAPOI	198296	11	18027	5.4679	0.000000
CONDITIO*DATAPOI	125683	22	5713	1.5933	0.040722

CoP x (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.023047	2	0.011523	0.47404	0.624026
DATAPOI	0.027594	11	0.002509	2.94041	0.000879
CONDITIO*DATAPOI	0.093001	22	0.004227	3.84441	0.000000

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIT	0.002178	2	0.001089	0.177548	0.837613
DATAPOI	0.000739	11	0.000067	0.666187	0.770898
CONDITIT*DATAPOI	0.002608	22	0.000119	0.930670	0.553676

Hip shoulder task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	400.94	2	200.471	3.50186	0.034311
DATAPOI	7234.55	11	657.686	8.67832	0.000000
CONDITIO*DATAPOI	4811.35	22	218.698	3.64185	0.000000

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	12.618	2	6.30910	0.383211	0.682776
DATAPOI	306.679	11	27.87990	2.933673	0.000902
CONDITIO*DATAPOI	195.927	22	8.90577	1.142317	0.293883

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	316947	2	158474	120.5587	0.000000
DATAPOI	173517	11	15774	18.8153	0.000000
CONDITIO*DATAPOI	113003	22	5136	6.3644	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.039242	2	0.019621	2.97142	0.056279
DATAPOI	0.039860	11	0.003624	3.91797	0.000018
CONDITIO*DATAPOI	0.029794	22	0.001354	2.77267	0.000024

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.087123	2	0.043562	5.874148	0.004001
DATAPOI	0.001320	11	0.000120	2.003179	0.026354
CONDITIO*DATAPOI	0.000771	22	0.000035	0.637495	0.899095

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	57.44	2	28.720	0.45525	0.635739
DATAPOI	15752.39	11	1432.035	14.46532	0.000000
CONDITIO*DATAPOI	5624.98	22	255.681	2.86334	0.000013

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	4.982	2	2.49123	0.209162	0.811657
DATAPOI	115.094	11	10.46312	1.287986	0.227766
CONDITIO*DATAPOI	56.976	22	2.58983	0.285532	0.999548

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	312563	2	156281	0.4427	0.643712
DATAPOI	207173	11	18834	44.2502	0.000000
CONDITIO*DATAPOI	79293	22	3604	7.9992	0.000000

CoP x (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.099043	2	0.049522	7.9523	0.000660
DATAPOI	0.182233	11	0.016567	13.4531	0.000000
CONDITIO*DATAPOI	0.078048	22	0.003548	4.2655	0.000000

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.066996	2	0.033498	5.240285	0.007034
DATAPOI	0.001042	11	0.000095	1.430352	0.155467
CONDITIO*DATAPOI	0.000727	22	0.000033	0.422441	0.991344

Hip twist task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	317.22	2	158.6092	8.511672	0.000412
DATAPOI	934.29	11	84.9354	3.736132	0.000038
CONDITIO*DATAPOI	787.61	22	35.8006	1.525599	0.057267

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITOI	1743.05	2	871.525	17.56267	0.000000
DATAPOI	5905.09	11	536.827	10.16579	0.000000
CONDITOI*DATAPOI	4784.70	22	217.486	3.80647	0.000000

Fz (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	338559	2	169280	132.737	0.000000
DATAPIN	216110	11	19646	32.402	0.000000
CONDITIO*DATAPIN	48874	22	2222	7.391	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.004048	2	0.002024	0.29605	0.744474
DATAPOI	0.023874	11	0.002170	9.91730	0.000000
CONDITIO*DATAPOI	0.015355	22	0.000698	3.01772	0.000004

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.011294	2	0.005647	0.48088	0.619821
DATAPOI	0.066944	11	0.006086	11.21265	0.000000
CONDITIO*DATAPOI	0.047644	22	0.002166	3.66169	0.000000

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	379.44	2	189.7212	8.423967	0.000443
DATAPOI	1478.53	11	134.4123	9.763482	0.000000
CONDITIO*DATAPOI	527.74	22	23.9881	1.473455	0.073766

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	812.49	2	406.2461	8.231004	0.000521
DATAPOI	1853.15	11	168.4679	2.716506	0.002056
CONDITIO*DATAPOI	2327.73	22	105.8061	1.504450	0.063525

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	340598	2	170299	103.471	0.000000
DATAPOI	212775	11	19343	28.025	0.000000
CONDITIO*DATAPOI	43938	22	1997	5.190	0.000000

CoP x (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.080397	2	0.040198	6.10100	0.003275
DATAPOI	0.017539	11	0.001594	14.65459	0.000000
CONDITIO*DATAPOI	0.015567	22	0.000708	4.70212	0.000000

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.165659	2	0.082829	6.29518	0.002761
DATAPOI	0.058662	11	0.005333	6.59146	0.000000
CONDITIO*DATAPOI	0.025396	22	0.001154	1.45207	0.081632

Hip reach task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	141.48	2	70.74113	4.036161	0.020957
DATAPOI	91.77	11	8.34260	0.826372	0.613657
CONDITIO*DATAPOI	94.87	22	4.31226	0.418905	0.991830

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	2896.1	2	1448.043	5.628067	0.004977
DATAPOI	3040.7	11	276.425	1.417643	0.161046
CONDITIO*DATAPOI	6233.3	22	283.333	1.231263	0.211507

Fz (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	217857	2	108928	30.0296	0.000000
DATAPOI	249327	11	22666	26.1642	0.000000
CONDITIO*DATAPOI	70064	22	3185	5.8999	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.023773	2	0.011887	0.37436	0.688793
DATAPOI	0.072861	11	0.006624	4.92723	0.000000
CONDITIO*DATAPOI	0.026593	22	0.001209	0.52921	0.963430

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.003570	2	0.001785	0.247029	0.781646
DATAPOI	0.000838	11	0.000076	0.988573	0.455576
CONDITIO*DATAPOI	0.001152	22	0.000052	0.763737	0.772730

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITOI	93.9	2	46.9719	0.182006	0.833903
DATAPOI	7877.8	11	716.1663	2.798728	0.001508
CONDITOI*DATAPOI	5087.3	22	231.2407	0.868976	0.637547

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	10.76	2	5.3801	0.196686	0.821800
DATAPIN	332.30	11	30.2092	1.926218	0.034051
CONDITIO*DATAPIN	196.84	22	8.9471	0.574367	0.941638

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	140002	2	70001	15.0030	0.000002
DATAPOI	350808	11	31892	24.6866	0.000000
CONDITIO*DATAPOI	137863	22	6266	5.7137	0.000000

CoP x (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.060550	2	0.030275	1.68197	0.191800
DATAPOI	0.077293	11	0.007027	3.40434	0.000144
CONDITIO*DATAPOI	0.096058	22	0.004366	1.23055	0.212091

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
CONDITIO	0.010653	2	0.005327	0.650421	0.524261
DATAPOI	0.000805	11	0.000073	0.762950	0.677317
CONDITIO*DATAPOI	0.002334	22	0.000106	1.073609	0.369792

MOVEMENT COMPONENT

Hip knee task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	21.44	2	10.721	2.48748	0.088819
DATAPOI	22266.70	11	2024.245	32.72640	0.000000
LOADS*DATAPOI	960.02	22	43.637	3.91794	0.000000

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1.129	2	0.56472	0.98181	0.378609
DATAPOI	173.300	11	15.75457	5.87942	0.000000
LOADS*DATAPOI	45.455	22	2.06612	2.22731	0.000988

Fz (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	50770.0	2	25385.0	2.42319	0.094402
DATAPOI	202993.3	11	18453.9	48.89933	0.000000
LOADS*DATAPOI	28104.8	22	1277.5	15.23877	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.026669	2	0.013334	13.15838	0.000010
DATAPOI	0.452513	11	0.041138	69.96974	0.000000
LOADS*DATAPOI	0.004699	22	0.000214	2.89983	0.000010

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.000220	2	0.000110	0.29574	0.744698
DATAPOI	0.000699	11	0.000064	2.26913	0.010528
LOADS*DATAPOI	0.000675	22	0.000031	4.05182	0.000000

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	15.68	2	7.842	1.89060	0.156929
DATAPOI	7336.05	11	666.914	22.83910	0.000000
LOADS*DATAPOI	1101.75	22	50.080	6.59522	0.000000

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.806	2	0.40311	0.760078	0.470608
DATAPOI	168.156	11	15.28690	4.687918	0.000001
LOADS*DATAPOI	17.062	22	0.77555	0.806987	0.719329

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	38184.2	2	19092.10	1.86515	0.160809
DATAPOI	215072.3	11	19552.03	23.36158	0.000000
LOADS*DATAPOI	19727.2	22	896.69	4.29086	0.000000

CoP x (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.019596	2	0.009798	8.29646	0.000493
DATAPOI	0.094948	11	0.008632	22.92672	0.000000
LOADS*DATAPOI	0.004116	22	0.000187	3.01259	0.000004

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.000660	2	0.000330	1.11875	0.331187
DATAPOI	0.002001	11	0.000182	6.24356	0.000000
LOADS*DATAPOI	0.000098	22	0.000004	0.63761	0.899006

Hip shoulder task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	101.533	2	50.766	12.1103	0.000022
DATAPOI	7268.877	11	660.807	37.6335	0.000000
LOADS*DATAPOI	819.911	22	37.269	7.5222	0.000000

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	5.6283	2	2.81416	5.80242	0.004263
DATAPOI	44.6994	11	4.06358	3.14842	0.000393
LOADS*DATAPOI	30.1105	22	1.36866	2.41166	0.000293

Fz (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	15028.6	2	7514.31	115.9152	0.000000
DATAPOI	95690.8	11	8699.17	42.1261	0.000000
LOADS*DATAPOI	19858.8	22	902.67	15.0863	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.051111	2	0.025556	31.4531	0.000000
DATAPOI	0.041325	11	0.003757	15.2359	0.000000
LOADS*DATAPOI	0.004612	22	0.000210	4.0538	0.000000

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.001105	2	0.000552	1.424243	0.246063
DATAPOI	0.000318	11	0.000029	2.431887	0.005878
LOADS*DATAPOI	0.000128	22	0.000006	1.556244	0.049159

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	3.83	2	1.917	0.50266	0.606608
DATAPOI	13293.03	11	1208.457	50.26983	0.000000
LOADS*DATAPOI	1518.88	22	69.040	10.53534	0.000000

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.6327	2	0.31636	1.09713	0.338250
DATAPOI	29.6454	11	2.69504	1.83905	0.045253
LOADS*DATAPOI	11.4659	22	0.52118	0.92554	0.560643

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	14694.7	2	7347.3	0.76674	0.467534

DATAPOI	115152.6	11	10468.4	80.94524	0.000000
LOADS*DATAPOI	21092.8	22	958.8	30.82363	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.022961	2	0.011481	19.0544	0.000000
DATAPOI	0.249720	11	0.022702	60.8431	0.000000
LOADS*DATAPOI	0.018511	22	0.000841	11.7555	0.000000

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.000482	2	0.000241	0.816632	0.445165
DATAPOI	0.000112	11	0.000010	0.568043	0.855066
LOADS*DATAPOI	0.000110	22	0.000005	1.066054	0.378737

Hip twist task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	15.117	2	7.55831	2.36459	0.099803
DATAPOI	1041.197	11	94.65430	14.64542	0.000000
LOADS*DATAPOI	107.414	22	4.88245	3.22821	0.000001

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	166.28	2	83.141	24.98117	0.000000
DATAPOI	3820.13	11	347.285	13.22735	0.000000
LOADS*DATAPOI	700.25	22	31.829	9.03210	0.000000

Fz (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	17520.8	2	8760.4	84.7532	0.000000
DATAPOI	118401.5	11	10763.8	60.8889	0.000000
LOADS*DATAPOI	20089.0	22	913.1	27.3129	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.014096	2	0.007048	16.54915	0.000001
DATAPOI	0.052407	11	0.004764	53.99958	0.000000
LOADS*DATAPOI	0.002652	22	0.000121	7.78219	0.000000

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.105549	2	0.052774	34.2383	0.000000
DATAPOI	0.221722	11	0.020157	49.2852	0.000000
LOADS*DATAPOI	0.007746	22	0.000352	9.9059	0.000000

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	105.594	2	52.797	12.89209	0.000012
DATAPOI	488.614	11	44.419	15.66406	0.000000
LOADS*DATAPOI	138.393	22	6.291	6.25936	0.000000

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	58.80	2	29.3991	7.31086	0.001143
DATAPOI	6965.73	11	633.2484	22.57386	0.000000
LOADS*DATAPOI	245.21	22	11.1459	2.65094	0.000057

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	17536.1	2	8768.1	71.8480	0.000000
DATAPOI	145123.5	11	13193.0	60.4185	0.000000
LOADS*DATAPOI	19557.1	22	889.0	22.5738	0.000000

CoP x (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.003338	2	0.001669	4.86353	0.009870
DATAPOI	0.027802	11	0.002527	52.84047	0.000000
LOADS*DATAPOI	0.002142	22	0.000097	11.27364	0.000000

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.043630	2	0.021815	17.63427	0.000000
DATPOIN	0.263808	11	0.023983	52.16073	0.000000
LOADS*DATPOIN	0.005950	22	0.000270	5.22952	0.000000

Hip reach task

Fx (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	167.32	2	83.658	0.41047	0.664576
DATAPOI	23162.05	11	2105.641	38.97888	0.000000
LOADS*DATAPOI	341.90	22	15.541	1.16174	0.274319

Fy (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	1.5701	2	0.7851	0.84120	0.434553
DATAPOI	32.6002	11	2.9637	1.80971	0.049726
LOADS*DATAPOI	9.4112	22	0.4278	0.64663	0.891739

Fz (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	25358.3	2	12679.2	64.7248	0.000000
DATAPOI	125918.1	11	11447.1	51.1909	0.000000
LOADS*DATAPOI	23492.9	22	1067.9	21.5147	0.000000

CoP x (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.013726	2	0.006863	3.8378	0.025151
DATAPOI	1.223599	11	0.111236	170.7913	0.000000
LOADS*DATAPOI	0.005956	22	0.000271	2.5287	0.000132

CoP y (condition 1)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.000348	2	0.000174	0.67180	0.513329
DATAPOI	0.001251	11	0.000114	7.10036	0.000000
LOADS*DATAPOI	0.000111	22	0.000005	1.07609	0.366873

Fx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	7612.81	2	3806.405	16.55456	0.000001
DATPOIN	4887.72	11	444.339	12.17076	0.000000
LOADS*DATPOIN	18420.90	22	837.314	28.87065	0.000000

Fy (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	36.784	2	18.39202	10.07008	0.000113
DATAPOI	79.163	11	7.19660	3.76376	0.000034
LOADS*DATAPOI	65.237	22	2.96530	2.25166	0.000844

Fz (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	48245.9	2	24123.0	70.9965	0.000000
DATAPOI	227809.5	11	20710.0	53.1484	0.000000
LOADS*DATAPOI	57807.5	22	2627.6	27.2060	0.000000

CoPx (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.245196	2	0.122598	25.8409	0.000000
DATAPOI	0.104784	11	0.009526	29.4521	0.000000
LOADS*DATAPOI	0.813194	22	0.036963	85.6010	0.000000

CoP y (condition 2)

	SS	Degr. of - Freedom	MS	F	p
LOADS	0.002731	2	0.001366	5.114032	0.007878
DATAPOI	0.000509	11	0.000046	3.583905	0.000070
LOADS*DATAPOI	0.000764	22	0.000035	3.360383	0.000000

BODY DISCOMFORT

Static body discomfort Tukey results

Tukey HSD test; variable DV_1 (body discomfort static 2.sta)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 2.1653, df = 282.00

Cell	TASKS	LOADS	{1} .72917	{2} .83333	{3} 1.5833	{4} .37500	{5} .47917	{6} 1.3958	{7} .39583	{8} 1.3333	{9} 2.7500	{10} 1.2292	{11} 1.9583	{12} 4.4792
1	1	1		1.000000	0.161790	0.990561	0.999589	0.535226	0.994338	0.685756	0.000018	0.884268	0.002528	0.000018
2	1	2	1.000000		0.341623	0.933750	0.990561	0.776427	0.951934	0.884268	0.000018	0.977034	0.009830	0.000018
3	1	3	0.161790	0.341623		0.003351	0.012684	0.999976	0.004421	0.999589	0.005802	0.990561	0.984969	0.000018
4	2	1	0.990561	0.933750	0.003351		1.000000	0.033035	1.000000	0.063209	0.000018	0.161790	0.000025	0.000018
5	2	2	0.999589	0.990561	0.012684	1.000000		0.094101	1.000000	0.161790	0.000018	0.341623	0.000068	0.000018
6	2	3	0.535226	0.776427	0.999976	0.033035	0.094101		0.041289	1.000000	0.000414	0.999993	0.776427	0.000018
7	3	1	0.994338	0.951934	0.004421	1.000000	1.000000	0.041289		0.077403	0.000018	0.190901	0.000029	0.000018
8	3	2	0.685756	0.884268	0.999589	0.063209	0.161790	1.000000	0.077403		0.000163	1.000000	0.636691	0.000018
9	3	3	0.000018	0.000018	0.005802	0.000018	0.000018	0.000414	0.000018	0.000163		0.000042	0.259547	0.000018
10	4	1	0.884268	0.977034	0.990561	0.161790	0.341623	0.999993	0.190901	1.000000	0.000042		0.387079	0.000018
11	4	2	0.002528	0.009830	0.984969	0.000025	0.000068	0.776427	0.000029	0.636691	0.259547	0.387079		0.000018
12	4	3	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	0.000018	

Dynamic body discomfort Tukey results

Tukey HSD test; variable DV_1 (body discomfort dynamic2.sta)
Approximate Probabilities for Post Hoc Tests
Error: Within MSE = 2.0664, df = 282.00

Cel	TASKS	LOADS	{1} 1.4167	{2} 1.8333	{3} 3.2708	{4} .43750	{5} .75000	{6} 1.6667	{7} .35417	{8} 1.1250	{9} 2.6875	{10} 1.0833	{11} 1.7292	{12} 3.5625
1	1	1		0.959911	0.000018	0.040285	0.496770	0.999486	0.015459	0.997852	0.000909	0.993090	0.996035	0.000018
2	1	2	0.959911		0.000075	0.000137	0.011968	0.999991	0.000045	0.396268	0.136794	0.305127	1.000000	0.000018
3	1	3	0.000018	0.000075		0.000018	0.000018	0.000020	0.000018	0.000018	0.701857	0.000018	0.000026	0.997852
4	2	1	0.040285	0.000137	0.000018		0.996035	0.001694	1.000000	0.445660	0.000018	0.548830	0.000665	0.000018
5	2	2	0.496770	0.011968	0.000018	0.996035		0.076776	0.972556	0.981909	0.000018	0.993090	0.040285	0.000018
6	2	3	0.999486	0.999991	0.000020	0.001694	0.076776		0.000485	0.792361	0.025301	0.701857	1.000000	0.000018
7	3	1	0.015459	0.000045	0.000018	1.000000	0.972556	0.000485		0.264285	0.000018	0.349252	0.000187	0.000018
8	3	2	0.997852	0.396268	0.000018	0.445660	0.981909	0.792361	0.264285		0.000023	1.000000	0.652286	0.000018
9	3	3	0.000909	0.136794	0.701857	0.000018	0.000018	0.025301	0.000018	0.000023		0.000020	0.050307	0.113710
10	4	1	0.993090	0.305127	0.000018	0.548830	0.993090	0.701857	0.349252	1.000000	0.000020		0.548830	0.000018
11	4	2	0.996035	1.000000	0.000026	0.000665	0.040285	1.000000	0.000187	0.652286	0.050307	0.548830		0.000018
12	4	3	0.000018	0.000018	0.997852	0.000018	0.000018	0.000018	0.000018	0.000018	0.113710	0.000018	0.000018	

EXAMPLE OF ELECTROMYOGRAPHY PRINTOUT DURING TESTING